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Preface

INTRODUCTION

The field of medical physics and related technologies is developing rapidly. The last several decades have witnessed the introduction of pioneering methods such as computed tomography, magnetic resonance and molecular imaging, to name just a few. Usually such new methods are applied with relative ease in healthcare and are further modified according to their field of application. This is a dynamic development process and often the available information about the new medical equipment and methods is rather limited. The present Encyclopaedia of Medical Physics is a link in a series of collaborative projects addressing this issue.

Two of the preceding steps in this process were the EMERALD and EMIT projects (www.emerald2.co.uk), which developed training materials (e-books and image databases) to address the initial training of medical physicists. These included specific training tasks covering the fields of x-ray diagnostic radiology, nuclear medicine, radiotherapy, ultrasound imaging and MR imaging. These materials include the first ever e-books with image databases and paved the way for e-learning in the profession. These are now used in more than 60 countries around the world (the EMIT project received the inaugural award for education of the European Union – the Leonardo da Vinci Award at Maastricht, 2004).

Both EMERALD and EMIT held conferences, which assessed the materials and revealed the need for a free professional reference source linked to a multilingual dictionary of terms. This web tool was expected to quickly provide information for new, existing and old methods and equipment in medical physics. The tool (EMITEL) was specially made to support the education, training and CPD process in the profession.

EMITEL PROJECT

The initial idea for this project was developed during the period 2001–2005 and was further prepared for submission to the EU. At this stage, the project partnership included the core of the previous project partners – King’s College London (KCL) (contractor) and King’s College Hospital, University of Lund and Lund University Hospital, University of Florence, AM Studio Plovdiv and the International Organization for Medical Physics (IOMP). This was the first EU project of IOMP as an institution and paved the way for further international projects and funding. At the middle of the project, a new partner joined the partnership – the Abdus Salam International Centre for Theoretical Physics (ICTP, Trieste), which later held the EMITEL Conference in 2008.

The objective of the new pilot project, EMITEL (European Medical Imaging Technology e-Encyclopaedia for Lifelong Learning), was to develop an original e-learning tool, which will be used for lifelong learning of a wide spectrum of specialists in medical physics. In addition to the e-encyclopaedia a multilingual digital dictionary of terms was planned to be developed to cross-translate the terms in any of its languages.

Medical imaging was emphasized in the name of the project, as this technology expands rapidly. However, radiotherapy and radiation protection were also included, together with a number of general terms associated with medical physics.

The EMITEL project (2006–2009) was funded by the EU programme, Leonardo da Vinci, as well as by the project partners. The EMITEL consortium has an agreement to continue its function after the end of the project, assisting the update of the project results.

EMITEL DICTIONARY

The EMITEL Dictionary used as a base the previous EMIT dictionary (initially available from www.emitdictionary.co.uk). The list of medical physics terms in this early dictionary was further refined and expanded. Currently, nearly 3400 terms are included. These terms were translated into 29 languages by colleagues from these countries. Thus, the original 7 languages – English, Swedish, Italian, French, German, Portuguese, Spanish – were supplemented by 22 new languages: Arabic, Bengali, Bulgarian, Chinese, Croatian, Czech, Estonian, Finnish, Greek, Hungarian, Iranian, Japanese, Korean, Latvian, Lithuanian, Malaysian, Polish, Romanian, Russian, Slovenian, Thai and Turkish.

The medical physics dictionary uses synchronized lists of terms, allowing cross-translation of terms between each two of its languages. The dictionary database is expandable and can include additional languages. The dictionary immediately became popular and currently has more than 2000 users every month (many from the developing countries, where limited professional literature is available in the specific languages).

The preparation of the dictionary was coordinated by S. Tabakov, and its software was developed by AM Studio (M Stoeva and A Cvetkov). The EMITEL consortium extends special gratitude to all translators (listed further below), who performed this task voluntarily.

EMITEL ENCYCLOPAEDIA

Each term from the dictionary is explained through an article (entry) in English. The entries, which vary from 50 to 500 words, aim at MSc level and above. The model of the encyclopaedia is built around a large number of specific entries (rather than a small number of multi-page articles), which allows for quick search and easy update. However, most of the EMITEL entries include further reading and information about other related entries in EMITEL, thus forming information strings.

Many of the entries include images, graphs, examples and other additional information. Very often this additional information is related to the images from the previous projects, EMERALD and EMIT. The consortium expresses its gratitude to all colleagues and companies who provided images and diagrams.

The encyclopaedic entries are grouped in seven categories – Physics of X-ray Diagnostic Radiology, Nuclear Medicine, Radiotherapy, Magnetic Resonance Imaging, Ultrasound Imaging, Radiation Protection and General Terms. Most entries include contributions from three specialists – author, referee and group coordinator.

A number of entries were written in parallel by several groups (e.g. Absorption) aiming to explain the entry from different perspectives. Some repetitions were allowed in order to facilitate the reader of a particular article. Where possible, abbreviations of terms (e.g. AEC) were included in the text as links to the respective full entries.
An original EMITEL website was created by AM Studio. The website uses the ability of the current Internet browsers to operate in all languages and combines the dictionary and the encyclopaedia. In this way, each translated term comes with a hyperlink displaying the corresponding entry. A multilingual search engine works with all languages of the dictionary.

The rapid development of medical physics has led to the existence of a number of acronyms and synonyms (included in the text). To deal with this problem, a second search engine was added to the website, which looks inside the full text of the entries (in English) and displays those entries where a particular synonym/acronym is mentioned. Care was taken, where possible, to include term modifications and variations.

The EMITEL website (www.emitel2.eu) is hosted by a web company. Alongside the database of terms, it has an additional internal website with content management system (CMS, also developed by AM Studio). The function of the CMS is to allow updates to existing entries and to add new information, images and diagrams. The CMS also allows for new terms to be added with new entries. In this way, EMITEL will act as the professional wikipedia of medical physics, with the difference being that only accepted entries and text will be uploaded (i.e. with editorial control). Currently, the EMITEL e-encyclopaedia has approximately 8000 users every month.

The present two volumes include the printout of all EMITEL entries (as in the website in 2012).

**EMITEL REFEREERING**

All entries of the encyclopaedia were refereed by specialists, who later became members of the team. To ensure that the entries are suitable for postgraduate-level usage, these were re-checked by MSc students from the MSc Medical Engineering and Physics, King’s College London and King’s College Hospital. Further text and images were included after this stage. The multistage refereeing of all entries was coordinated by the EMITEL administrator V. Tabakova, from the Division Imaging Sciences and Biomedical Engineering, KCL. This process took place in parallel with the completion of the encyclopaedia (2008) and continued until the end of 2011. All EMITEL referees are included in the list of contributors.

The intention of EMITEL is to continue the updates of the online material. In this way, contributors and referees are welcome to submit new articles, or to update existing articles.

**EMITEL NETWORK**

EMITEL is the largest international project in the profession. To develop and maintain the large volume of information in the encyclopaedia and dictionary, an international network was created. Currently it includes approximately 320 colleagues from 36 countries, more than half being translators. The network includes many senior professionals – officers and committee chairs of IOMP, EFOMP, AFOMP, AAPM, IPEM, IFMBE and current and past presidents of over 20 national medical physics societies.

The network was first discussed and agreed during the EMITEL International Conference on Medical Physics, held at ICTP, Trieste, 24–26 October 2008 (ICTP – The Abdus Salam International Centre for Theoretical Physics). The conference was attended by colleagues from 22 countries, who evaluated the materials and decided on the way forward.

The conference delegates included the current IOMP president, secretary general, treasurer, chair of ETC, chair of AHC, IFMBE secretary general, IUPESM secretary general, EFOMP president-elect, AFOMP president, AAPM president, IPEM vice-president and many distinguished colleagues from Europe and the rest of the world.

Currently, the internal links of the network are maintained through an administrator in KCL. This function will gradually transfer to the IOMP. It is expected that the number of network members will increase, as in the future colleagues who contribute new articles and materials to EMITEL will also be included. In this way, the network will assure the future support and expansion of the EMITEL encyclopaedia and dictionary as a website free to use by all colleagues.

**ACKNOWLEDGEMENTS**

EMITEL gratefully acknowledges the financial support from the EU Leonardo Programme and the support from the partner institutions. EMITEL is most grateful for the professional input and materials delivered by its many contributors, forming the EMITEL network. EMITEL gratefully acknowledges all materials of companies and colleagues used to enhance the articles.

Slavik Tabakov  
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President-Elect of IOMP
How to Use EMITEL Website (www.emitel2.eu)

Select Encyclopaedia > write the term you want to see at the win-
dow > click Enter. A list with terms is displayed – against each
one is a blue hyperlink related to the area of the term > click the
hyperlink to read the article. EMITEL can also search inside
the text of the articles. To do this, select Search in Full Text;
after this, specify the area and proceed as mentioned earlier. In
case of UK or American English differences (i.e. colour>color;
optimise>optimize) try both spellings or search only part of the
term (e.g. colo, optim).

To use the Dictionary select Dictionary > choose the input and
output languages > write the term you want to see at the window
> click Enter. A list with terms is displayed, where the terms are
found either single, or in combination with other words (the e-Dic-
tionary assumes that the user’s Internet browser already supports
the input and output languages).

To use both the encyclopaedia + dictionary select combined and
proceed as mentioned earlier (this search is limited only to the title
of the article, not inside its text).

The website also includes new entries, updates, errata and other
corrections, which have been submitted by colleagues and approved
by the Editorial Board.

DISCLAIMER
The EMITEL Encyclopaedia and Dictionary include text, images
and diagrams. These have been checked for viruses. EMITEL can
be read with any Internet browser. The e-Dictionary assumes that
the Internet browser already supports the input language and output
languages.

All efforts were made for the correct translation of terms in the
dictionary. Similarly, the articles in the EMITEL encyclopaedia
have been subject to internal and external refereeing. However, the
EMITEL Consortium would welcome additional suggestions for
their improvement.

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2D arrays (Ultrasound) 2D arrays are transducers in which there are multiple rows of transducer elements in order to provide improved control of beam width in the elevation plane or to provide beam steering in the elevation plane. In the latter case, 2D arrays allow volume scanning of tissue without the need for moving parts, and these are available commercially for cardiac and abdominal imaging. The different geometries and nomenclatures are outlined in the article on Matrix array.

Related Article: Matrix arrays

3D (three-dimensional) (General) 3D is an abbreviation for three dimensional. This refers to a dimensional space built up by three arbitrary dimensions, e.g. length, width and depth. The position of any point in a 3D space can be described using three coordinates in the Cartesian coordinate system.

Related Article: Cartesian coordinate system

3D imaging (Nuclear Medicine) This is the process of acquiring data representing the three-dimensional radioactive source distribution in an object. A common approach used in nuclear medicine imaging is to acquire many planar images from different angles from which to produce a 3D image dataset. This technique is called tomographic imaging. The 3D distribution is reconstructed from the planar images in a post-processing procedure. Tomographic imaging techniques in nuclear medicine include single photon emission computed tomography (SPECT) and positron emission tomography (PET).

Related Articles: Positron emission tomography (PET), Single photon emission computed tomography (SPECT), Image reconstruction

90° pulse (Magnetic Resonance) In MRI pulse sequences an arbitrary RF pulse can be applied to flip the bulk magnetisation vector through some angle. 90° RF pulses are implemented very frequently in pulse sequence designs, utilised both as inverting pulses and refocusing pulses.

As an inverting pulse, the function of a 180° pulse is to invert longitudinal magnetisation (Figure 1.2). Subsequent to the inverting pulse the longitudinal magnetisation starts to recover to its equilibrium state through T1 relaxation. After a delay, tissues with different T1 values will have recovered to different degrees. STIR and FLAIR type sequences exploit the longitudinal magnetisation differences that emerge after an inverting pulse to null signal from fat and fluid, respectively. Equally, a preparatory, 180° inversion pulse can be utilised to enhance T1 weighting in a ‘host’ imaging sequence. Sequences utilising preparatory 180° pulses are generally referred to as ‘inversion recovery type sequences’.

180° pulses are also implemented in spin echo type sequences to ‘refocus’ magnetisation and generate a spin echo. After an excitation pulse, magnetisation is flipped onto the transverse (xy) plane. Individual spin transverse components dephase (Figure 1.3) in the transverse plane due to local variations in precession frequencies determined by T2* mechanisms. The result is a ‘fan’ of spin components, moving apart from one another, or dephasing. The application of a 180° pulse (Figure 1.3) moves the slow, trailing edge of the fan of spins ahead in phase relative to the fast, leading edge. As the fan now closes, the spin rephase and the signal builds up generating a spin echo.

**FIGURE 1.1** 90° pulse showing flipping of longitudinal magnetisation onto the transverse plane.

**FIGURE 1.2** 180° inverting pulse.

**FIGURE 1.3** Refocusing using a 180° pulse.
A number
(Nuclear Medicine) See Mass number

AAPM TG43 formalism
(Radiotherapy, Brachytherapy) The source models recommended today (2009) for brachytherapy treatment planning systems are those based on the AAPM TG43 formalism.

The TG43 formalism has water as the reference dosimetry medium. Reference data for commercially available sources are compiled from Monte Carlo calculations and experimental data. (Reference data are available from AAPM and ESTRO.)

Only cylindrical sources are considered, and cylindrical symmetry is assumed. A reference point is defined 1 cm from the source centre on the transverse bisector of the source, \((r_0, \theta_0)\) with \(r_0 = 1\) cm and \(\theta_0 = \pi/2\) (polar co-ordinates).

The dose rate to water in water at point \(P(r, \theta)\) is given by Equation A.1:

\[
D(r, \theta) = S_K \frac{G(r, \theta)}{G(r_0, \theta_0)} \frac{g(r)}{g(r_0)} F(r, \theta) \quad (A.1)
\]

The AAPM TG43 formalism, dose rate to water in water at point \(P(r, \theta)\)

- \(S_K\) is the air kerma strength in units \(U\), \(1\) \(U = 1\) \(\mu\)Gy m\(^2\)/h = 1 Gy cm\(^2\)/h.

- \(A\) is the dose rate constant in water, the dose rate to water in water at the reference point per unit air kerma strength. The constant \(A\) must be determined for all source types and designs (unit cGy/h/U).

- \(G(r, \theta)\) is the geometry function describing the distribution of the radioactive material in the source (proportional to \(1/r^2\) for a point source).

- \(g(r)\) is the radial dose function, describing the dependence of absorption and scatter of the photons in water along the transversal bisector \((\theta_0 = \pi/2)\), also including the effect of interactions in the source material and encapsulation. The function \(g(r)\) is dimensionless and normalised at 1 cm, \(g(r_0) = 1\).

- \(F(r, \theta)\) is the anisotropy function describing the anisotropy of the dose distribution around the source ‘in relation to the transversal bisector plane’, the absorption and scatter of photons in the source itself, in the encapsulation and in the water. The function \(F(r, \theta)\) is dimensionless and normalised along the transversal bisector, \(F(r_0, \theta_0) = 1\).

Care must be taken when dose rate tables are used. The tables to be used, e.g. as data in a treatment planning system must be derived using the same formalism/definitions as the formalism used in that specific system.

For a detailed description of the TG43 formalism, please refer to the references given in the following.

**Abbreviations:** AAPM = American Association of Physicists in Medicine, ESTRO = European Society for Therapeutic Radiology and Oncology.

**Related Articles:** Treatment planning systems – Brachytherapy, Source models


**Abdominal imaging**
(Diagnostic Radiology) The physical parameters of anatomical structures have specific effects on the use of various imaging modalities. Abdominal imaging is difficult, as the anatomy of the region includes solid and hollow structures which complicate detection of lesions (compared with chest imaging). Imaging of these structures requires the use of specific contrast media. The most commonly used intravenous contrast agents (x-ray imaging) are iodine based. Barium-based contrast agents (x-ray imaging) are usually used orally (e.g. barium meal to diagnose the hollow gastro-intestinal tract). The high subject contrast in these cases requires the use of higher energies (kVp). The most commonly used modalities for abdominal imaging are computed tomography (CT) and magnetic resonance imaging (MRI), while ultrasound is very useful in the rare cases of imaging during pregnancy.

**Related Article:** Chest radiography

**Abdominal imaging**
(Nuclear Medicine) Abdominal imaging is a generic term for medical imaging of abdominal disorders, including the alimentary tract and the genitourinary system. It may include diagnostic ultrasound imaging (US), CT, single photon emission tomography (SPECT), positron emission tomography (PET), and MRI.

There are many nuclear medicine studies involving abdominal imaging. A few examples of these are Tc-99 m colloid for liver imaging, Tc-99 m Imidodiacetates for biliary function studies, Tc-99 m labelled red cells for GI bleeding, In-111 labelled white cells for infection and Ga-67 and I-123 MIBG for tumour imaging.

**Abdominal imaging**
(Ultrasound) Abdominal imaging in ultrasound includes assessment of the liver, gall bladder, kidneys and renal tract, pancreas, spleen, retroperitoneum and gastrointestinal tract and surrounding structures. B-mode, colour flow and spectral Doppler are all used in abdominal imaging, usually with curvilinear transducers in the range 1–6 MHz for adults.
Aberration

(Ultrasound) In optics, aberration refers to the failure of light rays to converge at a single focus. The analogous problem in ultrasound imaging is that a focused beam will not produce a sharp focus, and conversely, that a point reflector will appear blurred in the image.

The first basic prerequisite for ultrasound imaging to work at all is that echoes are produced, which occurs when there is a transition in acoustic impedance (the product of sound speed and density) between two types of tissues. Second, there should ideally be no difference in sound speed along the path of interrogation, as the distance to the interface is given by the time of the returning echoes and an assumption of the sound speed. Clearly, these conditions are contradictory as tissue is heterogeneous, but since the sound speed variations between tissue types are relatively small, ultrasound imaging works well to a first order.

**Aberration Correction:** In ultrasound imaging, the beamformer phases and sums signals from individual transducer elements. Ideally these signals differ only in the arrival time, which relates directly to the distance between the receiving element and the reflector. However, the propagation in tissue breaks up the amplitude consistency, and makes also parts of the wavefront arrive earlier or later than had the wave propagated in a homogenous medium. This is due to the combined effects of scattering, reverberation, refraction and the cumulative differences in time delay associated with the passage through tissue layers of different types and thicknesses. Studies of aberration correction aim to correct for these tissue propagation effects, to recover the full potential of the imaging system.

**Correction Procedures:** A variety of correction methods have been devised to solve this problem, usually applied to sites like the abdominal wall and the chest wall. Usually, two steps are involved: determining the degree of aberration and then to correct for it in an adaptive way. In astronomy, atmospheric aberration can be corrected by maximising the intensity integral in the image plane through real-time adjustments of time delay. In ultrasound imaging of the body, point targets (like stars in astronomy) are not readily available, so these methods have limited applications, except for certain cases like kidney stones. However, random backscatter (as from tissue) can also provide measuremen of the phase error. The usual approach is therefore to assume a phase screen in front of the transducer, i.e. to assume that the main aberration effect is a phasing effect. By cross correlation of signals on adjacent elements, an estimate of the phase error can be obtained. Once this is known, a delay of the opposite sign is applied to compensate for the aberration. Iteration can reduce the error further. To make an optimal correction, investigators have shown that also amplitude aberration can be corrected by maximising the intensity integral through real-time adjustments of time delay.

At an atomic level the variations in specific interactions between incident ionising radiation and individual atoms or molecules in an absorber implies that the resultant distribution of energy absorption is a stochastic process. The definition of absorbed dose takes this into account by averaging out such variations over a volume per unit mass of absorber, and can therefore be considered a macroscopic quantity.

Absorbed dose is therefore formally defined as the mean energy absorbed within a mass of absorber:

$$D = \frac{\overline{dE}}{dm}$$

where $\overline{dE}$ is the mean energy imparted in joules (J) over a mass of tissue $dm$.

**Related Articles:** Mean absorbed dose in air, Mean energy imparted, Kerma, Air kerma

**Absorbed dose conversion factor**

(Radiation Protection) The absorbed dose conversion factor, $f_{X,D}$, is the relation between the absorbed dose in air and the exposure. That is,

$$D = f_{X,D} \times X$$

where

- $D$ is the absorbed dose in air
- $X$ is the exposure

Even if $X$ is only well defined in air, sometimes, especially for diagnostic radiology or nuclear medicine energies, where electronic equilibrium is easily established, it is possible to obtain an acceptable approximation for absorbed dose in other media such as water or muscle tissue using tabulated values of absorbed dose conversion factors for these media.

In a similar way, a dose conversion factor, $f_{K,D}$, can be defined to calculate absorbed dose from air kerma measurements:

$$D = f_{K,D} \times K$$

where

- $D$ is the absorbed dose
- $K$ is the kerma

At higher energies, for instance in radiotherapy, absorbed dose conversion factors are usually associated to chamber calibration factors.

**Related Articles:** Exposure, Kerma, Absorbed dose

**Absorbed dose distribution**

(Radiation Protection) Absorbed dose distribution is a description, often a graphic description, of energy deposition across the matter where the ionising radiation impinges on. Due to the lack of homogeneity in the medium and to the radiation attenuation, the energy
deposition may adopt different patterns, which may be more or less complex.

A usual way to represent the absorbed dose distribution is to use isodose curves.

Sometimes absorbed dose distribution may be determined using a set of dosimeters (such as TLDs) located in the real medium or in a simulated one. In many cases absorbed dose distributions must be calculated by means of an appropriate mathematical model or through a Monte Carlo simulation

Related Articles: Absorbed dose, Energy deposition

**Absorbed fraction**

*(Nuclear Medicine)* When determining the radiation dose received by a target organ from a source organ, the final step is to determine the absorbed fraction ‘φ’. The absorbed fraction is a measure of the fraction of the energy emitted from a source organ that is absorbed in the target organ. The absorbed fraction depends on the target organ composition and the amount of radiation reaching it (i.e. the distance between the two organs). For example, a high absorbed fraction indicates that the source and target organ are located adjacent or near to each other and that the target organ is ‘composed’ of a high attenuating tissue. The absorbed fraction is therefore calculated for each emission type and source-target organ pair. The notation \( \phi_i (r_k \leftarrow r_h) \) (Gy kg) indicates the absorbed fraction from a source organ \( r_k \) to the target organ \( r_h \) from the \( i \)th emission. Thus the dose absorbed by the target organ is given by

\[
D(r_h \leftarrow r_k) = \frac{\bar{A}}{m_t} \sum_i \phi_i (r_k \leftarrow r_h) \Delta_i \quad (A.2)
\]

where

- \( D \) (Gy) is the total dose to the target organ received from the source organ
- \( \bar{A} \) (MBq s) is the cumulated activity
- \( m_t \) (kg) is the target organ mass
- \( \Delta_i \) (Gy kg/MBq s) is the equilibrium absorbed constant for the \( i \)th emission

The non-penetrating radiation (photons with energy less than 10keV and electrons) is assumed to deposit all its energy locally, i.e. in the source organ.

Values of the absorbed fraction are calculated using a phantom model of the human body. The phantom which has been used for many years is one developed by the MIRD committee. The phantom is a standard model of the human being with fixed organ sizes and anatomic relationships. Even though the phantom is a relatively good approximation of the average human, the deviation from the standard phantom can be large for specific patients.

**Specific Absorbed Fraction:** The specific absorbed fraction \( \Phi \) is given by the quotient between the absorbed fraction and the target organ mass *(Equation A.3)*:

\[
\Phi = \frac{\phi}{m_t} \quad (A.3)
\]

\( \phi \) is the fraction of radiation emitted by the source organ that is absorbed per unit mass in the target organ mass and has the unit Gy. The absorbed dose to the target organ using the specific absorbed fraction (insertion of Equation A.3 in Equation A.2) is given by

\[
D(r_h \leftarrow r_k) = \bar{A} \sum_i \Phi_i (r_k \leftarrow r_h) \Delta_i \quad (A.4)
\]

**Dose Reciprocity Theorem:** Consider an organ pair. The theorem states that the specific absorbed fraction is the same, regardless of which organ is the target and which is the source, i.e. the energy absorption per gram is identical regardless whether the radiation is travelling between \( r_k \) and \( r_h \) or \( r_h \) and \( r_k \). This theorem is useful when the absorbed fraction is not available for all source–target pairs. For example, when \( \phi_i (r_k \leftarrow r_h) \) is known, then \( \phi_i (r_h \leftarrow r_k) \) can be calculated according to

\[
\frac{\phi(r_h \leftarrow r_k)}{m_h} = \frac{\phi(r_k \leftarrow r_h)}{m_k} \quad (A.5)
\]

The absorbed fraction and the equilibrium absorbed dose constant are often combined into a mean dose per cumulated activity \( S \) value (see separate article) to simplify the calculation procedure. The \( S \) value is determined by the emission type, radiation energy and anatomic relationship and is determined for each source–target pair and radionuclide.

Related Articles: Cumulated activity, Equilibrium absorbed dose constant, MIRD formalism, Mean dose per cumulated activity


**Absorbed radiation** *(Radiation Protection)* When radiation goes through matter, a variety of phenomena may happen. Those phenomena can be classified in three classes: a number of photons or particles pass through without interacting; some others may be scattered with or without loss of energy; and finally a part of the incident radiation delivers all of its energy to the matter. This is the absorbed radiation.

At low energies, absorbed radiation is mainly associated with the photoelectric effect. Pair production also contributes to radiation absorption at higher energies.

Related Articles: Scattered radiation, Photoelectric absorption, Photoelectric effect, Pair production

**Absorber, broad-beam geometry** *(Radiation Protection)* The geometry used for an x-ray exposure will effect the measured transmission of the x-ray beam, and the subsequently calculated attenuation on an absorber. Broad-beam and narrow-beam geometry set-ups are shown in Figure A.1a and b.

In the broad-beam set-up shown in Figure A.1a, radiation that would have reached the detector (A) is scattered and does not reach the detector. However, some radiation that would not originally have reached the detector is scattered so that it does (B). This may lead to an increase in detector reading. For narrow-beam geometries (Figure A.1b), it is assumed that the primary x-ray beam has been collimated and therefore any scattered radiation will miss the detector.

The result of this difference in set-ups is that for a broad-beam geometry, the same absorber does not appear to attenuate as much as if it were part of a narrow-beam geometry set-up. As a result, the value of the half-value thickness calculated for the absorber would be different for each set-up. This effect has practical implications, e.g. the field size must be reduced when x-raying patients. If a broader beam than necessary is used, extra scatter reaches the detector and contrast is reduced.

It is important to note that for a certain absorber, the radiation absorbed will be the same for both narrow- and broad-beam
Absorber density

(set-ups. It is the scattering effects that cause the apparent decrease in attenuation for broad-beam geometries compared to narrow-beam geometries.


Absorber density

(Radiation Protection) The density of a particular absorber is dependant on its mass and volume (Equation A.6). The more dense the absorber, the more likely photon interactions are to occur. An increase in density will lead to an increase in absorption and scatter, and therefore an increase in attenuation. Such increased absorption in denser materials implies that they receive a higher radiation absorbed dose:

\[ \rho = \frac{m}{v} \]  

(A.6)

where

- \( \rho \) is the density (kg/m³)
- \( m \) is the mass (kg)
- \( v \) is the volume (m³)

Absorber, linear attenuation coefficient

(Radiation Protection) Attenuation of photonic radiation is an exponential process that can be represented by the equation

\[ I = I_0 e^{-\mu x} \]

where

- \( I_0 \) is the initial beam intensity
- \( I \) is the final beam intensity
- \( x \) is the distance travelled
- \( \mu \) is the attenuation coefficient of the material (absorber)

The coefficient \( \mu \) normally refers to the linear attenuation and describes the attenuation property of an absorber in terms of attenuation per unit thickness. The unit for \( \mu \) is 1/length (e.g. cm⁻¹).

See Attenuation.

Related Articles: Attenuation, Mass attenuation coefficient, Atomic attenuation coefficient

Absorber, mass attenuation coefficient

(Radiation Protection) The amount of attenuation in an absorber may be related to the mass or density of the absorbing material in terms of its mass attenuation coefficient, given by

\[ \frac{\mu}{\rho} \]

where

- \( \mu \) is the linear attenuation coefficient
- \( \rho \) is the density of the material

For more information, see article on Attenuation.

Related Articles: Attenuation, Mass attenuation coefficient

Absorber, mean free path for photons

(Radiation Protection) Mean free path for photons in absorbers is shorter than that for photons in gas due to the increased density of molecules within the absorber. This increased density is analogous with an increased value of \( n \) in Equation A.7:

\[ l = \frac{1}{n\pi r^2} \]  

(A.7)

where

- \( l \) is the mean free path
- \( n \) is the number of molecules
- \( r \) is the collision radius of the molecule

Related Article: Mean free path

Absorber, narrow-beam geometry

(Radiation Protection) The geometry of the x-ray set-up may affect the output measured by the detector. This is described in the article Absorber, broad-beam geometry.

Related Articles: Absorber, broad-beam geometry

Absorption

(Radiation Protection) Absorption is the process by which the energy of a beam of radiation incident on a material is imparted to the atoms or molecules of the material. Some of the energy in the incident beam will possibly be transmitted out of the material. Any material that absorbs energy in this way is called an absorber. Materials can also scatter the incident radiation to different directions through various processes (Compton scatter, elastic scatter, etc.). The scattered radiation may also be absorbed or may be transmitted out of the material.

The most important absorption process at diagnostic x-ray energies is the photoelectric effect, and is the main mechanism by which the energy of the radiation is transferred to the material, potentially leading to damage if the material is human tissue made up of biological molecules, cells and most importantly, DNA.

Related Articles: Photoelectric effect, Compton scatter

Absorption

(Ultrasound) The intensity of a propagating ultrasound wave is decreased by distance. This phenomenon is called attenuation and is due to scattering, reflection, divergence and absorption of the ultrasound beam energy. Absorption is caused when there is imperfect relationship between the pressure changes in the sound wave and corresponding resulting density changes in the medium. The lost mechanical energy is converted into heat.
The sound energy is lost and cannot be recovered. Absorption is dependent on tissue composition and structure and increases with frequency.

**Related Articles:** Attenuation, Damping, Intensity, Scattering

### Absorption coefficients

*(Radiation Protection)* For a given medium, the absorption coefficient, \(\mu_a\), is a measure of its photon radiation absorption efficiency. When a beam of intensity \(I_0\) traverses a medium of thickness \(x\) the amount of radiation absorbed in it is given by

\[
I = I_0 \times (1 - \exp(-\mu_a \times x))
\]

Absorption coefficients depend on the medium composition, material density and radiation energy. Material composition dependence is determined by the atomic number of the medium or by the so-called effective atomic number when a mixture of more than element is present. Density dependence may be avoided through the use of mass absorption coefficients.

Radiation absorption may arise from several interaction processes. So, several absorption coefficients can be defined, for instance, associated with photoelectric absorption, \(\mu_{a,pe}\), and with pair production, \(\mu_{a,pp}\). The total absorption coefficient is the combination of all of them:

\[
\mu_a = \mu_{a,pe} + \mu_{a,pp} + \cdots
\]

In general, the attenuation of radiation is the result of absorption and scattering, so the absorption coefficients combine with scattering coefficients as well.

**Related Articles:** Mass absorption coefficients, Scattering coefficients, Attenuation coefficients

### Absorption cross section

*(Ultrasound)* The absorption cross section is defined as the time-averaged total absorbed power divided by the time-averaged incident intensity. The unit is in square meters. Physically this cross section corresponds to the area of the incident wave that contains the amount of power that is absorbed by an object. This means that the absorption cross section divided by the geometrical cross section of the object is a measure of how effectively the object absorbs sound. See also Scattering cross section, Extinction cross section and Differential scattering cross section.

### Absorption efficiency

*(Radiation Protection)* Absorption efficiency is a term used to describe how efficiently a detector absorbs incident ionising radiation. The term might be used either in the field of Radiation protection (see Absorbed dose) and associated measurements, or in the field of diagnostic radiology imaging (see DQE).

**Related Articles:** Absorbed dose, Detective quantum efficiency.

### Absorptive backing

*(Ultrasound)* Absorptive backing is used in an ultrasound transducer to damp out vibrations in the transducer element in order to produce short pulses (Figure A.2).

To produce short pulses the backing material should be highly absorptive and have acoustic impedance close to that of the piezoelectric material. However, this requirement runs counter to that of maximising the ultrasound output and therefore the amount of damping is a compromise between sensitivity and pulse length. In certain applications, such as in pulsed or continuous wave, sensitivity is more important than spatial resolution and a less absorptive material can be used.

**Related Articles:** Damping, Attenuation, Absorption, Matching layer, Bandwidth

### AC generator

*(General)* An alternating current generator is based on turning a conductor coil in a magnetic field (Figure A.3).

The movement of the rotating coil causes the magnetic flux through the coil to vary sinusoidally as the coil rotates. This generates a sinusoidally varying electrical potential or which can be used as a source of electrical power. The frequency of the AC power is directly related to the rotational speed of the rotor.

The sketch shows a single coil and slip ring pair, though mounting three coils and slip ring sets at 120° intervals on the rotor would enable the generation of three-phase power.

### AC motor

*(General)* An AC motor is a motor that is powered by alternating current. The motor is made up of an outer stationary part (the stator) which produces a magnetic field and an inner rotating part (the rotor) which produces an opposing field that generates a reactionary rotational force.

There are many types of AC motor designs, though they can be considered in two basic categories: the induction and the synchronous motor.

Synchronous motor – the rotor turns synchronously with the changing phases of the AC current, and the rotor is either a coil powered through slip rings or can be a permanent magnet. One example is the shaded pole motor used in electric clocks.

Induction motor – here the rotor is not powered but has current induced in it by the changing magnetic field provided by the stator.

**Related Articles:** Damping, Attenuation, Absorption, Matching layer, Bandwidth

**Related Articles:** Good axial resolution, Short pulse length, Magnetic field, Sinusoidal alternating voltage

**FIGURE A.2** The effect of backing a transducer element. (Graphs courtesy of EMIT project, www.emerald2.eu)

**FIGURE A.3** AC generator principle.
This type rotates slightly slower than synchronously. Common forms include the ubiquitous ‘squirrel cage’ rotor and the more expensive ‘wound’ rotor. These are found in most domestic equipment. Very large, powerful, electric motors are usually three-phase induction motors.

**Related Article**: AC generator


### Accelerating waveguide

**(Radiotherapy)** See Wave guide

### Acceleration compensation

**(Magnetic Resonance)** During an ordinary MR imaging sequence, the object is expected to be at rest. If this is not the case, and the object moves during and/or between data acquisition, artefacts such as mis-positioning and blurring may occur in the reconstructed images.

Both moving and static spins will accumulate a phase offset when exposed to a gradient. However, the phase offset induced on the moving spin is different from the phase of the static spin. Additionally, if the spin is accelerating or jerks, the spin phase will have an additional phase contribution. The different phase angle results in mis-positioning and blurring of the object in the Fourier reconstruction.

Flow compensation can be done by introducing additional gradient lobes prior to the echo readout. The aim of these lobes is to null the phase offset of the spins induced by first-order motion (velocity). Likewise, by adding additional gradient prior to readout, second-order (acceleration) induced phase offsets and even effects of jerk can also be nullled, although at the prize of the utilisation of rather complicated gradient lobe patterns and prolonged echo time.

**Related Articles**: Flow compensation, Phase contrast, Velocity mapping

### Accelerator

**(General)** Accelerators use electric fields to accelerate charged particles to high velocity and energy. There are two main types of accelerators used in medical physics.

Firstly linear accelerators in radiotherapy accelerate electrons in a straight line to produce either a high-energy photon beam (by collision with a target) or an electron beam for treatment. Their design and use are explained in more detail in the article *Linear accelerator*.

Secondly, cyclotrons are used in the production of radioisotopes in nuclear medicine, notably PET imaging. A cyclotron accelerates a beam of charged particles (protons, deuterons, or alpha particles) along a circular path to gain energy and then bombard targets of certain elements. This process produces positron-emitting radioisotopes such as $^11$C, $^13$N, $^15$O and $^18$F. This is explained further in the article *Cyclotron*.

**Related Articles**: Linear accelerator, Cyclotron

### Accelerator-produced radionuclides

**(Nuclear Medicine)** Radionuclides are mainly produced in two ways, either in a reactor or in a cyclotron. Charged particles are accelerated in a cyclotron and directed towards a target; hence, the radionuclides produced in a cyclotron are referred to as *accelerator-produced radionuclides*. An alternative approach is to use photon-induced reactions by irradiating a target with high-energy photons (∼100MeV). A downside with using photons is that the most common nuclear interaction for low Z elements is the $(\gamma, n)$. For example, $^{18}$F is produced by irradiating $^{19}$F-gas. Since it is almost impossible to separate two isotopes of the same element one can never obtain carrier-free $^{18}$F with this approach. Carrier-free radionuclides are acquired when using a cyclotron to irradiate $^{18}$O with a proton beam, giving a $(p, n)$ reaction. Therefore cyclotron production is preferred to photon-induced production.

Tables A.1 and A.2 contain clinically relevant accelerator-produced photon and positron emitters.

**Related Articles**: Carrier-free sample, Collimator, Cyclotron, Positron emission tomography


<table>
<thead>
<tr>
<th>Radionuclide</th>
<th>$T_n$</th>
<th>Mode Off Decay (%)</th>
<th>Main $\gamma$-Ray Energy (keV) (%)</th>
<th>Nuclear Reaction</th>
<th>Energy Range (MeV)</th>
<th>Thick Target Yield MBq(mCi)/µA h</th>
</tr>
</thead>
<tbody>
<tr>
<td>$^{65}$Ga</td>
<td>3.26 d</td>
<td>EC (100)</td>
<td>93 (37) 185 (20) (199.6)</td>
<td>$^{65}$Zn $(p,2n)$</td>
<td>26 → 18</td>
<td>185 (5)</td>
</tr>
<tr>
<td>$^{111}$In</td>
<td>2.8 d</td>
<td>EC (100)</td>
<td>173 (91) 247 (94)</td>
<td>$^{111}$Cd $(p, 2n)$</td>
<td>25 → 18</td>
<td>166 (4.5)</td>
</tr>
<tr>
<td>$^{123}$I</td>
<td>13.2 h</td>
<td>EC (100)</td>
<td>159 (83)</td>
<td>$^{123}$Te $(p,n)$</td>
<td>14.5 → 10</td>
<td>137 (3.7)</td>
</tr>
<tr>
<td>$^{123}$Xe</td>
<td>$^{123}$Te $(p,2n)$</td>
<td>26 → 23</td>
<td>392 (10.6)</td>
<td>$^{123}$I $(p,5n)$</td>
<td>65 → 45</td>
<td>777 (21)$^a$</td>
</tr>
<tr>
<td>$^{123}$Xe</td>
<td>$^{123}$Te $(p,2n)$</td>
<td>29 → 23</td>
<td>414 (11.2)$^b$</td>
<td>$^{123}$Xe $(p,3n)$</td>
<td>28 → 20</td>
<td>18 (0.5)$^d$</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Production Data</th>
<th>Nuclear Reaction</th>
<th>Energy Range (MeV)</th>
<th>Thick Target Yield MBq(mCi)/µA h</th>
</tr>
</thead>
</table>

These radionuclides are primarily used for scintillation camera and SPECT imaging.

$^a$ $^{123}$Xe decays to $^{123}$I by EC (87%) and $\beta^-$ (13%) emission.

$^b$ $^{123}$I yield expected from $^{123}$Xe decay over an time of approximately 7h.

$^c$ $^{203}$Pb decays to $^{203}$TI by EC (100%).

$^d$ $^{203}$TI yield expected from the decay of $^{203}$Pb after 32h.
**Accidental coincidences of PET systems**

Accidental coincidences refer to false coincidences, i.e., the decay has occurred outside the line of response. Examples of accidental coincidences in PET are scattered, spurious and random coincidences. To read more about each individual event type, read the related articles.

**Related Articles:** Object scatter events, Scatter coincidence, Random coincidence, True coincidence, Spurious coincidence, Event type in PET

**Accumulator (storage battery)**

An accumulator or storage battery is a device that converts chemical energy into electrical energy, consisting of a group of electrochemical cells that are connected to act as a source of direct current. The cells are encased in a container and fitted with terminals to provide a source of direct electric current at a given voltage. A cell consists of two dissimilar substances, a positive electrode and a negative electrode, that conduct electricity, and a third substance, an electrolyte, that acts chemically on the electrodes. The two electrodes are connected by an external circuit; the electrolyte functions as an ionic conductor for the transfer of the electrons between the electrodes. Batteries are classed as either dry cell or wet cell. In a dry cell the electrolyte is absorbed in a porous medium, or is otherwise restrained from flowing. In a wet cell the electrolyte is in liquid form and free to flow and move. Batteries also can be generally divided into two main types – rechargeable and nonrechargeable, or disposable. Disposable batteries, also called primary cells, can be used until the chemical changes that induce the electrical current supply are complete. Rechargeable batteries, also called secondary cells, can be reused after being drained. This is done by applying an external electrical current, which causes the chemical changes that occur in use to be reversed. A battery called a storage battery is generally of the wet-cell type, i.e., it uses a liquid electrolyte and can be recharged many times. The storage battery consists of several cells connected in series. Each cell contains a number of alternately positive and negative plates separated by the liquid electrolyte. The positive plates of the cell are connected to form the positive electrode; similarly, the negative plates form the negative electrode. In the process of charging, the cell is made to operate in reverse of its discharging operation; i.e., current is forced through the cell in the opposite direction, causing the reverse of the chemical reaction that ordinarily takes place during discharge, so that electrical energy is converted into stored chemical energy. Batteries are made of a wide variety of electrodes and electrolytes where the most common are lead and sulphuric acid, alkaline battery (alkaline), nickel cadmium (NiCd), nickel hydrogen (NiH2), nickel metal hydrate (NiMH), lithium ion (Li-ion), and lithium ion polymer (Li-ion polymer).

**Accuracy**

The term accuracy is used in quality control assessments. It describes the ability of a system to keep its parameters exactly as set up.

For example, x-ray tube kVp accuracy refers to the ability of the x-ray system (the generator and the tube) to produce exposures with accurate kVp (the measured kVp to be identical with the set...
Acetic acid in film processing

(Diagnostic Radiology) Acetic acid has been used as a stop bath in film processing. The purpose of the stop bath is to quickly stop the development process before the film is placed in the fixer. A stop bath is not used in the processing of radiographic films. Rollers in the film transport system squeeze some of the developer solution out of the emulsion and help stop the developer activity.

**Related Article:** Film processing

Acoustic axis (Ultrasound) The ultrasonic beam axis is defined as the line fitted to points of maximum acoustic pressure measured at increasing distances in the direction of propagation of a transmitted ultrasound field.


Acoustic impedance (Ultrasound) The quantity characteristic acoustic impedance ($Z$) is defined as the ratio of the driving force (acoustic pressure, $p$) to the response (local particle velocity, $v$) and is a measure of how difficult it is for a particle to move within a medium. This relation is analogues to Ohm’s law where the electrical impedance is the ratio of the voltage (driving force) to the current (response). The acoustic impedance can also be expressed as the product of the medium’s density ($\rho$) and its speed of sound ($c$). For a plane wave with linear propagation, $Z$ can be obtained by

$$Z = \frac{p}{v} = \rho c$$

The dimensions of acoustic impedance are $M/L^2/T$. The units are Rayls, where 1 Rayl = 1 Pa s/m

Values of the acoustic impedance for some common materials and types of human tissue are shown in the table:

<table>
<thead>
<tr>
<th>Material</th>
<th>$Z$ (kg/m$^2$/s$^2$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Water</td>
<td>$1.48 \times 10^6$</td>
</tr>
<tr>
<td>Air</td>
<td>430</td>
</tr>
<tr>
<td>Blood</td>
<td>$1.67 \times 10^6$</td>
</tr>
<tr>
<td>Muscle</td>
<td>$1.71 \times 10^6$</td>
</tr>
<tr>
<td>Skull bone</td>
<td>$6.47 \times 10^6$</td>
</tr>
</tbody>
</table>

The acoustic impedance is a very important tissue parameter in diagnostic ultrasound as reflection of pressure waves occur at boundaries with materials with different values of $Z$, see Reflection coefficient.

**Related Articles:** Speed of sound, Reflection coefficient

Acoustic power (Ultrasound) Acoustic power is a measure of the rate at which an ultrasound transducer produces acoustic energy. Acoustic power is measured in watts (W). Typical values of acoustic power for diagnostic scanners are in the order of 10–100 mW. However, in some Doppler modes higher values can be reached. Acoustic power is a most important parameter when calculating the risk of heating tissue.

The output power is defined as time-averaged ultrasonic power radiated by an ultrasonic transducer into an approximately free field under specified conditions in a specified medium, preferably water. Acoustic power $W$ can also be expressed as the integration of intensity $I$ over a specific area $S$:

$$W = \int I dS$$

Acoustic power is best measured using a force balance.

**Related Articles:** Force balance, Radiation force, Intensity, Thermal index


Acoustic pressure (Ultrasound) Acoustic pressure describes the pressure perturbation during the passage of a sound wave, as opposed to the static pressures (such as atmospheric and hydrostatic). The acoustic pressure is the pressure component detected by a hydrophone or ultrasound transducer. In common derivations of the wave equation, it is assumed that the acoustic pressure variations are small compared to the static pressure.

**Acoustic streaming (Ultrasound)** Acoustic streaming is a sound wave- or vibration-induced time-independent flow. It results from a transfer of momentum to the liquid as it absorbs acoustic energy. There are three main types of acoustic streaming:

1. The small-scale streaming that occurs near viscous boundary layers to an oscillating object. In this type of streaming the vortices are much smaller than the wavelength of the sound wave from the oscillating object. These types of flow can be seen in Figure A.4.

2. The medium-sized acoustic streaming occurs outside the boundary layer of the oscillating object. The streaming is also rotational in nature but the size of the vortices is in the same scale as the wavelength. These types of vortices are also shown in Figure A.4.
The final type of acoustic streaming is the largest where the vortices are much bigger than the wavelength. This type of streaming occurs when a beam of sound propagates in a volume of liquid larger than the beam itself. Figure A.5 depicts such a streaming. As can be seen the volume of liquid determines the size of the vortices.

**Related Articles:** Acoustic power, Safety

**Acoustic working frequency**

(*Ultrasound*) The acoustic working frequency of an acoustic signal is based on the observation of the output of a hydrophone placed in an acoustic field at the position corresponding to the spatial-peak temporal-peak acoustic pressure. The signal is analysed either using the zero-crossing frequency technique or using a spectrum analysis method.

When analysing a pressure spectrum the acoustic working frequency is the arithmetic mean of the most widely separated frequencies $f_1$ and $f_2$ at which the amplitude of the pressure spectrum of the acoustical signal is 3 dB lower than the peak amplitude (Figure A.6).

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Unit</th>
</tr>
</thead>
<tbody>
<tr>
<td>$f_{acf}$</td>
<td>MHz</td>
</tr>
</tbody>
</table>


**Acoustics**

(*Ultrasound*) Acoustics is the scientific study of the production and properties of sound waves. The word ‘acoustic’ often refers to the entire frequency range while the word ‘sound’ is divided in infrasound (0–20 Hz), sound (20–20 kHz, human hearing interval) and ultrasound (>20 kHz).

The Greek word ‘acoustic’ means ‘able to be heard’ and ‘sonic’ is the Latin synonym.

**Acquisition modes for digital image**

(*Diagnostic Radiology*) Imaging modalities use various modes for acquiring digital images. The ‘acquisition mode’ for digital images was a term used at the beginning of digital medical imaging when systems had only a limited number of acquisition modes. For example, the fluoroscopic image from an image intensifier was acquired (and digitised) in two forms – interlaced and progressive mode. The interlaced mode (the one used in broadcast TV) displays first the odd lines of the TV raster, then the even lines. This way the digital image is formed by two combined half-frames. The reason for using the interlaced mode is that it prevents perceived flicker in the viewed image because it fills the full display from top to bottom faster. Alternatively the progressive (non-interlaced mode) displays all lines of each frame in sequence. Hence the final digital image is formed by one frame that contains all of the lines. This mode however requires not only a change in the TV raster function (with doubled frequency), but also a more powerful analogue to digital converter. Another mode, not used today as a term, was the frame mode, which acquired separate images (frames) with certain speed (frames per second) – a mode similar to cut film changer operation.

**Acquisition time**

(*Nuclear Medicine*) The acquisition time refers to the time span where the detector system counts and registers events. The length of the acquisition time is primarily determined by the detector count rates. For example, in order to attain an acceptable signal to noise ratio at low count rates, a long acquisition time is required.

When imaging organs with sequential movements, like a beating heart and a breathing lung, the spatial resolution is severely degraded by motion distortions when using a continuous acquisition. An alternative is to use gated acquisition. The acquisition time is divided into small portions of the total sequence time. Each acquisition time will be triggered at a certain point in the sequence, hence registering events only during specific parts of the organ motion sequence.

**Related Article:** Gated acquisition
Acrylic is known chemically as polymethyl methacrylate (PMMA) and by trade names such as Perspex and Plexiglass. It is a transparent thermoplastic, a synthetic polymer of methyl methacrylate. Figure A.7 shows its chemical structure. It was developed in 1928 and released commercially in 1933 by Rohm and Haas Company.

PMMA is usually used as an alternative to glass, because of its low cost and machine-ability. Its density is less than half that of glass, similar to other plastics. It is brittle when loaded and softer than glass (i.e. easily scratched). A 3 mm thickness transmits up to 98% of visible light entering through its surface, and reflects around 4% of incident light from each of its surfaces. It has good environmental stability compared to other plastics, but has a poor resistance to solvents.

**Medical Applications:** PMMA has a good degree of compatibility with human tissue; therefore it can be used in a variety of medical applications. It is used as a replacement intraocular lens in the eye when the original lens has been removed in the treatment of cataract. In orthopaedics, PMMA bone cement is used to affix implants and to remodel lost bone. Dentures are often made of cataract. In orthopaedics, PMMA bone cement is used to affix implants and to remodel lost bone. Dentures are often made of PMMA, and can be colour-matched to the patient’s teeth and gum tissue.

PMMA is also extensively used in medical physics applications, predominantly for quality assurance tests. It acts as a cheap alternative to specially developed phantoms to simulate patient conditions. It acts as a cheap alternative to specially developed phantoms to simulate patient conditions. This can therefore be regarded as constant.

**Acrylic**

(General)

<table>
<thead>
<tr>
<th>Property</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Molar mass</td>
<td>15–500 kg/mol</td>
</tr>
<tr>
<td>Density at STP</td>
<td>1150–1190 kg/m</td>
</tr>
<tr>
<td>Melting point</td>
<td>403–553 °C</td>
</tr>
<tr>
<td>Boiling point</td>
<td>473 °C</td>
</tr>
<tr>
<td>Refractive index</td>
<td>1.4893–1.4899</td>
</tr>
<tr>
<td>CT number</td>
<td>110–130 HU</td>
</tr>
</tbody>
</table>

**Figure A.7** Chemical structure of the PMMA polymer (C₆H₄O).  

**Acrylic (General)**

**Activation cross section**

**Radiation Protection** When linear accelerators (linacs) are operated above energies of 10 MeV significant radioactivity may be induced in the metal components in the head of the linac as a result of photon and neutron activation. Figure A.8 shows that at 18 MV the main radionuclides produced are from such n,γ reactions.

*The likelihood of this activation process occurring is called the activation cross section, and depends on the type and energy of radiation being produced by the linac, other output parameters such as the monitor units (MU) and total beam-on time, and the atomic number of the components within the linac head.*

Other articles such as furnishings and fittings within the treatment room may also have radioactivity induced in them by activation. Similarly the air in the room may also become slightly radioactive due to γ n reactions with nitrogen and oxygen, producing positron emitters with a short half-life.

**Activation formulas**

<table>
<thead>
<tr>
<th>Radionuclide</th>
<th>Half-Life</th>
<th>Probable Nuclear Reaction</th>
<th>Decay Mode</th>
<th>Principal Gamma-Ray Energies (MeV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Al-28</td>
<td>2.3 μs</td>
<td>27Al(n,γ)28Al</td>
<td>β, γ</td>
<td>1.78</td>
</tr>
<tr>
<td>Mn-56</td>
<td>2.6 μs</td>
<td>54Mn(n,γ)55Mn</td>
<td>β, γ</td>
<td>0.85, 1.81, 2.11</td>
</tr>
<tr>
<td>Na-24</td>
<td>15.0 μs</td>
<td>23Na(n,γ)24Na</td>
<td>β, γ</td>
<td>1.37, 2.75</td>
</tr>
<tr>
<td>Sb-122</td>
<td>2.8 d</td>
<td>125Sb(n,γ)125Sb</td>
<td>β, γ</td>
<td>0.51, 0.56</td>
</tr>
</tbody>
</table>

**Figure A.8** Principal activation products in the head of a linear accelerator operating at 18 MV. (Reproduced from NCRP Report No 151, Structural shielding design and evaluation for megavoltage X- and gamma-ray radiotherapy facilities, December 2005.)

**Related Articles:** Cross section, Neutron, Neutron activation cross section, Photon, Positron, Radionuclides


**Activation cross sections in radionuclide production**

**Nuclear Medicine** The amount of activity produced when a target is irradiated with particles is described in the activation formula and depends on the intensity of the particle beam, the number of target nuclei and the reaction probability. This reaction probability is expressed in terms of an effective area or a cross section σ. The Système International unit for σ is m² but it is commonly expressed in barn or millibarn. (1 mb = 10⁻³ b = 10⁻³¹ m².)

**Related Article:** Activation formula


**Activation formula**

**Nuclear Medicine** The amount of activity produced when a target is irradiated with particles depends on the intensity of the particle beam I, the number of target nuclei N₀ and the reaction probability, i.e. *activation cross sections* σ. (Please see articles on Activation formula thin target and Activation formula thick target.)

**Related Articles:** Activation formula thin target, Activation formula thick target


**Activation formula thick target**

**Nuclear Medicine** If a target is irradiated with particles it can lead to the production of radioactive species. The rate, R, at which this occurs depends on the intensity of the particle beam, the number of target nuclei and the reaction probability, i.e. activation cross sections. For typical irradiation durations and typical cross sections only a very small amount of the target material is consumed and this can therefore be regarded as constant.

If a thick target is irradiated, the incident particles will decrease in kinetic energy as they progress through the target (in contrast with a thin target where the kinetic energy is assumed to be constant). This will result in a variation of corresponding activation cross sections as these depend on the energy of the incident particle. To express the activation in a thick target the excitation function must be integrated. If we consider the thin target equation (please see article Activation formula thin target)
The solution to this equation reads

\[ R(t) = \frac{N}{\lambda} (1 - e^\lambda) \]

This gives us a radioactivity of

\[ A(t) = R \cdot \lambda = N(1 - e^\lambda) = \frac{I \cdot \rho \cdot N_A \cdot \sigma \cdot x}{M} (1 - e^\lambda) \]

This formula is only valid for constant cross sections, i.e. thin targets. For thick targets one needs to integrate over the entire energy range deposited in the target.

**Related Articles:** Activation formula thick target, Avogadro's number, Activation cross sections


**Activation rates in radionuclide production**

*(Nuclear Medicine)* See Activation formula

**Activators**

*(Diagnostic Radiology)* The primary function of the activator, (typically sodium carbonate), in film processing is to soften and swell the emulsion so that the reducers can reach the exposed grains. See also Accelerators.

**Related Article:** Film processing

**Active breathing control**

*(Radiotherapy)*

**Principle:** Active breathing control (ABC) involves controlling the patient’s respiratory cycle to reduce the effects of internal anatomy motion on the delivered dose distribution in external beam radiotherapy. A typical ABC approach is to use a mouthpiece with a flow sensor and valve. The flow sensor is interfaced to a control computer that monitors the patient’s breathing, producing a display showing the breathing pattern as a sine wave-like trace.

Several approaches exist to implement breathing control. Passive breathing control involves asking the patient to hold their breath using an audio cue or by adjusting their breathing to fit a desired pattern using the trace displayed on the computer screen. A typical active breathing control approach is to use a mouthpiece with a flow sensor and valve. The flow sensor is interfaced to a control computer that monitors the patient’s breathing, producing a display showing the breathing pattern as a sine wave-like trace. The valve is used to hold the patient’s breath at a desired position in the inhale or exhale part of breathing.

Figure A.9 shows an ABC system.

**Treatment Sites:** Treatment sites affected by motion due to breathing are the main candidates for the use of ABC. These include lung, liver and breast. ABC has also been used in the breast to exclude the heart from the radiation field to the left side of patients.

**Abbreviation:** ABC = Active breathing control, also Active breathing coordinator.

**Related Articles:** Deep inspiration breath hold, Gating – respiratory control
Active device

(Magnetic Resonance) An active device is any device that can only serve its intended use with an external supply of power by any means including electrical line, battery or gas power. Examples of active devices are ventilators, pacemakers, electroencephalographs, electrocardiographs and patient monitoring devices. The fringe magnetic field at a relative low strength may influence the functionality of an active device above a certain field strength which depends on the specific device. Implantable active devices include cardiac pacemakers, neurostimulators, cardiac defibrillators, drug infusion pumps, cochlear implants and insulin pumps. The severity of any effect of the magnetic field on the active device may vary from one device to another and there are also variations in the functionality thresholds for the same active devices. Some devices may be labelled MR safe or MR compatible but, this will only apply to the specific conditions stated in the manufacturer’s specifications.

Related Articles: MR safe, MR compatible

Active implant

(Magnetic Resonance) Active implants include any medical device that can only serve its intended use with the supply of power by any means including but not limited to line, battery or gas power. The use of an active implant is contraindicated for MR imaging because they are either magnetically, electrically or mechanically activated and this activation could be affected by exposure to the fields present in a magnetic resonance scanner. Therefore patients with such devices should not be examined with MR. These active implants include cardiac pacemakers, neurostimulators, implantable cardiac defibrillators, implantable drug infusion pumps, cochlear implants and insulin pumps. The use of electronically activated devices may also cause excessive heating that can result in burn injuries to patients undergoing MR procedures, as a result of conductive materials that have an elongated shape, such as electrodes, leads, guide wires and certain types of catheters (e.g. catheters with thermistors or other conducting components).

Related Article: Implant

Active matrix array

(Diagnostic Radiology) The active matrix array is an integrated circuit formed out of a large number of photodetector elements connected to thin film transistors (TFTs). It can be produced as a large area matrix (currently in excess of 40 × 40 cm²), which allows it to be used as a fundamental constituent in modern digital x-ray imaging detectors. For medical imaging the active matrix array is used for both direct and indirect radiography. Arrays used for both types of imaging incorporate a two-dimensional array of imaging pixels, which consists of a switching element used for data read-out (typically a thin film transistor, TFT) and a sensing and storage element.

The active matrix utilises thin film technology which allows the deposition of hydrogenated amorphous silicon (a-Si:H), making it ideal for construction of both TFTs and photodiodes. Large area arrays are formed by plasma deposition of thin layers of the appropriate materials (e.g. amorphous silicon) onto a glass substrate. Once deposited, they can be etched to the desired pattern by a process called photolithography.

Figure A.10 shows a typical array used in a medical imaging flat-panel detector. Within the array each pixel consists of a switching element and of an element able to detect incoming photons by storing them as charge. The image read-out process is controlled by altering the voltage applied across the switching element. Firstly, to allow each pixel to detect a signal during exposure, the voltage across each switching element is set to an ionisation or ‘off’ state. The signal is then read-out by changing the switching voltage row-by-row to the conducting or ‘on’ state which allows the charge stored in each pixel to be drained by the charge collector electrode and passed to the multiplexer. The voltage change is controlled by the gate line driver. As the read-out process is controlled by the external circuitry, each row of pixels requires a separate control line driver to alter the switching voltage, and each column its own amplifier. This process is called the active matrix read-out.

The active matrix array allows the radiographic image signal to be read-out sequentially, line by line. Fluoroscopic images are acquired in real-time by permitting all other rows that are not being read-out to continue to detect the incoming signal during exposure.

Abbreviations: AMA = Active matrix array, TFT = Thin filmed transistor.
Active matrix liquid crystal flat-panel display

(Diagnostic Radiology) An active matrix thin film transistor liquid crystal display is a thin, flat screened display whose pixels are created from an array of liquid crystal (LC) cells which are each individually controlled by a separate TFT in an active matrix array. In diagnostic radiology digital radiographs are usually viewed on an active matrix flat-panel TFT LC display. They have become widely popular in the medical imaging industry because they are smaller and lighter weight than their traditional cathode ray tube (CRT) counterparts, offering equivalent, if not superior resolution, contrast, viewing angle and response time.

The active matrix array is used to precisely control each individual pixel. Within a monochrome display each pixel consists of the switching element (usually a TFT) which controls a pixel electrode (made of indium tin oxide – ITO, transparent electrode) which in turn controls the transmission state of the LC and a storage capacitor to maintain a constant voltage throughout the LC. Figure A.11 shows a pixel cross section of a typical LC display.

The active matrix forms the image by controlling the pixel elements, row-by-row, shown in Figure A.12. Each row of pixels is selected by applying the appropriate select voltage to the gate line connecting all the TFT gates in one row. When a row of pixels is selected, the desired signal is applied to each pixel in that row via the data lines. In that way, each pixel is selected individually, and if the switching element works ideally, each pixel is addressed with no cross-talk between adjacent pixels.

For monochrome displays each pixel is formed from a single LC cell; in a coloured display each pixel is formed from three sub-pixels of red, blue and green, Figure A.13. To create a colour pixel a coloured filter is usually imbedded on the second glass substrate (Figure A.10); however, some development has been done to allow the filter to be integrated into the TFT substrate. The final coloured

Related Articles: TFT (thin film technology), Amorphous silicon detector


Active matrix liquid crystal flat-panel display

FIGURE A.11 The cross section of a typical TFT display pixel using a twisted nematic liquid crystal configuration.

FIGURE A.12 An active matrix array used for LCDs.

FIGURE A.13 (See colour insert.) Coloured display pixel configuration.
image is then created by altering the intensity and brightness of each coloured cell, using a similar principle to the coloured phosphor screens used in CRT displays.

**Related Articles:** TFT (thin film technology), Matrix array, Active matrix array, LCD (liquid crystal display), Viewing angle.

### Active shielding

**(Magnetic Resonance)** Active shielding refers to technical methods used in MRI scanner design to counter undesirable effects of both the static and gradient fields.

Ideally the static field should be contained completely inside the bore of the scanner. In real systems a fringe field will exist outside the scanner, the strength of which falls with distance. The fringe field presents a safety hazard due to both the missile effect on ferromagnetic objects and the potential for interference with the operation of electro-medical devices. In general, MRI installations are designed so that fringe fields no higher than five Gauss are present outside the scan room, or at least outside an area designated as a controlled zone. Prior to the introduction of active shielding MRI installations had to routinely incorporate ‘passive’ steel shielding in the scan room walls to contain the magnetic field. Passive shielding can add tens of tonnes to a room design and is a significant consideration in structural design. With active shielding a set of magnetic field coils in the MRI counters the field external to the machine. This reduces the footprint of the fringe field and the five gauss contour may be completely contained within a room of acceptably small dimensions without the need for passive shielding. Static field shielding is achieved with a set of superconducting coils external to the main static field superconductor windings. The shield produces a field opposing the field produced external to the MRI by the main field coil.

In gradient field switching, the resulting rapid changes in magnetic field strength induce local eddy currents in conductive parts of the MRI machine. These eddy currents generate magnetic fields which interfere with the applied gradient field, resulting in gradient field distortion. In relation to the bore of the magnet, the gradient field coils sit internal to the main field coils and cryostat. With active shielding of the gradient an intermediary set of shielding coils sits between the gradient coils and main field coils and cryostat. These active shielding coils are designed to counter the gradient field external to the gradient coils preventing eddy current induction in the scanner, while having minimal effect of the field internal to the gradient coils.


### Active transport of tracers

**(Nuclear Medicine)** Active transport of tracers refers to transport mechanisms that require energy. Actively transported substances are able to move against concentration gradients. Examples of active transport are the sodium–iodine pump and the renal tubular reabsorption of glucose.


### Activity

**(Nuclear Medicine)** The activity of a radionuclide is the average decay rate. It is measured in disintegrations per second. It is essentially a measure of how radioactive a substance is.

The average decay rate, $\Delta N/\Delta t$ in a sample containing $N$ atoms is determined by the decay constant $\lambda$. The decay constant is the probability for disintegrations of a single atom per second $(s^{-1})$. The mathematical expression is

$$\frac{\Delta N}{\Delta t} = -\lambda N$$

As the minus sign indicates, the number of atoms $N$ decreases with time. The activity $A$ refers to the number of disintegrations per second of $N$ radionuclides. The SI unit is the Becquerel [Bq] where a sample with activity of 1 Bq decays at a rate of 1 disintegration per second. The number of atoms $N$ at a specific time point $t$ depends on the number of atoms at $t = 0$ and the decay factor. The decay factor $e^{-\lambda t}$ is an exponential function that depends on the decay constant $\lambda$ and time $t$. The mathematical expression is

$$N(t) = N(0)e^{-\lambda t}$$

Since activity is proportional to the number of atoms, Equation A.9 can be translated into

$$A(t) = A(0)e^{-\lambda t}$$

Each radioactive nuclide is associated with a certain probability for decay. A high probability suggests that the atom is likely to decay and vice versa. When considering a greater number of atoms, it is more suitable to talk about the half-life $T_{1/2}$ of the sample; namely the time it takes for radioactivity to decrease to 50% of its initial activity level. The half-life and the decay constant are related according to

$$T_{1/2} = \frac{\ln 2}{\lambda}$$

In tables of radionuclides the half-life is often listed instead of the decay constant.

The activity of radionuclides used in Nuclear Medicine tends to be multiples of becquerels, e.g. kilobecquerels ($1\text{kBq} = 10^3$ disintegrations/s), megabecquerels ($1\text{MBq} = 10^6$ disintegrations/s) and gigabecquerels ($1\text{GBq} = 10^9$ disintegrations/s).

The curie is another unit of activity. One curie is equivalent to $3.7 \times 10^{10}$ disintegrations/s. One curie was originally defined as the activity of 1 g of $^{226}$Ra.


### Actual focal spot

**(Diagnostic Radiology)** See Focal spot actual

### Adaptive processing

**(Ultrasound)** Adaptive processing in ultrasound covers a wide range of techniques designed to enhance the ultrasound signals from tissue features while reducing artefactual signals in the image, particularly those from speckle. Adaptive processing algorithms permit changes in response to local changes in time and space and may be based on the statistical analysis of signals. A number of filtering techniques have been studied for ultrasound images including as median filters, Weiner filters and wavelets. Increased computational power has enabled the implementation of several...
Adaptive radiotherapy

Adaptive radiotherapy (ART), as its name suggests, involves delivering a radiotherapy treatment that is adapted to changes during the treatment course. Two basic concepts exist: adaptive planning and adaptive treatment delivery.

**Adaptive Planning:** The first approach developed for ART involved acquiring a set of planning scans of the patient on several imaging visits to get a measure of the variation in the anatomy expected from day to day, such that a target volume may be constructed based on an estimate of the expected inter-fraction variation for the patient’s anatomy. This was first demonstrated for the prostate using a set of five CT scans. The combined information from the scans may be used to generate a confidence limited planning target volume, cl-PTV. This enables a probabilistic approach to be used in treatment planning of target coverage.

**Adaptive radiotherapy delivery**

A newer approach to ART is to measure the patient’s anatomy and estimate the dose distribution during each treatment fraction, accumulate this information and plan subsequent treatment fractions to correct for discrepancies between planned and delivered treatment. In this process the dose delivery for subsequent treatment fractions of a course of radiotherapy can be modified to compensate for inaccuracies in dose delivery that cannot be corrected for by simply adjusting the patient’s positioning. The causes of these inaccuracies may include tumour shrinkage, patient weight loss and increased tumour hypoxia resulting during the course of fractionated treatment. Image-guided radiotherapy (IGRT) is a requirement for this technique.

**Abbreviations:** ART = Adaptive radiation therapy, cl-PTV = Confidence limited planning target volume, CT = Computed tomography, IGRT = Image guided radiotherapy and PTV = Planning target volume.

**Related Articles:** Target volume, Planning target volume


Adaptive responses and hormesis

(Radiation Protection) The current internationally accepted framework for radiation protection is based on a model of potential harm from exposure to ionising radiation called the linear no-threshold model. This model suggests that at any level of received radiation dose, harm may be caused – i.e. a cancer may be induced. The risk of harm (stochastic effects) is proportionate to the dose received (i.e. linear with dose). This is described in the diagram (Figure A.16):

However, more recent published evidence would suggest that this model is too simplistic and that for a number of reasons low-level exposure to ionising radiation may be more or less harmful than predicted by a linear extrapolation, and indeed may actually be beneficial. Models predicting less than expected damage to radiation exposure may be attributed to adaptive responses by the cell. If the radiation exposure provides benefit in terms of resistance to future carcinogenic events, then it is called hormesis. This hormetic effect appears in epidemiological evidence from mortality rates amongst British radiologists, survivors of Chernobyl, and studies in Taiwan. The diagram (Figure A.17) describes the possible consequence of hormesis to our understanding of the dose–response curve for stochastic effects.

**Figure A.14** Longitudinal image of a kidney without adaptive processing.

**Figure A.15** Longitudinal image of a kidney with XRES adaptive processing.

different adaptive processing methods in commercial systems. An example of Philips XRES system in 2007 is shown in Figures A.14 and A.15.

**Figure A.16** The linear no-threshold model.
The ATP molecule contains two phosphate groups, designated to a resonance peak at a different chemical shift in a \( ^{31}\text{P} \) spectrum. The molecule (Figure A.19). The phosphorus nucleus in each group and because most of the ADP (around 80%) is bound to proteins and hence is not NMR-visible. These resonances are generally not seen however, because they overlay the resonances due to the \( \gamma \) and \( \alpha \) phosphate groups in ATP and because most of the ADP (around 80%) is bound to proteins and hence is not NMR-visible.

The importance of ADP in \( ^{31}\text{P} \) NMR arises from its role in the body's energy metabolism. ADP is the product of dephosphorylation of ATP, the 'currency' of intracellular energy. Although ADP concentration usually cannot be directly measured using NMR, it can be estimated using the formula

\[
[\text{ADP}] = \frac{[\text{ATP}][\text{Cr}]}{K_f[\text{PCr}][\text{H}^+]}
\]

where \( K_f \) is the forward rate constant of the creatine kinase reaction.

Related Articles: ATP, Magnetic coupling

Adenosine triphosphate (ATP)

(Magnetic Resonance) ATP (adenosine triphosphate) is a chemical compound that features in in vivo phosphorus (\(^{31}\text{P} \)) NMR spectra. The ATP molecule contains three phosphate groups, designated \( \alpha \), \( \beta \) and \( \gamma \) in order of increasing distance from the remainder of the molecule (Figure A.20). The phosphorus nucleus in each group experiences a different electronic environment and hence gives rise to a resonance peak at a different chemical shift in a \( ^{31}\text{P} \) spectrum. These chemical shifts are sensitive to intracellular pH, Mg\(^{2+} \) concentration and temperature, and can be used to estimate these parameters.

Furthermore, the \( \gamma \) and \( \alpha \) resonances appear as doublets and the \( \beta \) resonance as a triplet due to homonuclear \( J \)-coupling
Adhesive, more commonly known as glue, is a compound that bonds two objects together. Adhesives come from either natural or synthetic sources. Natural or bio-adhesives are produced from inorganic mineral sources or biological sources including plant matter, starch, resins and animal tissues. Synthetic adhesives can be elastomers, thermoplastics and thermosets.

Adhesives can be further categorised depending on their method of adhesion. Drying adhesives usually are mixtures of polymers in a solvent, such as white glue and rubber cements. The adhesive hardens as the solvent evaporates. Drying adhesives are generally weak and tend to adhere to different materials to varying extents.

Contact adhesives are applied to both surfaces and are allowed to dry before the surfaces are put in contact, often requiring several hours to dry. Natural rubber and polychloroprene are common contact adhesives and are used in laminates and footwear.

Hot adhesives, such as the common 'glue gun', are thermoplastics which are applied when hot and harden on cooling.

Reactive adhesives function by either chemical bonding with the material's surface or hardening due to the polymerisation of two chemicals. Examples of such adhesives are two-part epoxy, silane, and metallic cross-links. They are used to prevent loosening of bolts in moving assemblies such as engines.

UV light curing adhesives or materials (LCMs) experience rapid curing, strong bonding and have the ability to withstand high temperatures. They are therefore suitable for the manufacture of products in the electronics, telecommunications, medical and aerospace industries. They are also used to seal and coat products.

Pressure-sensitive adhesives (PSAs) form a bond by the application of pressure due to a balance between flow and resistance to flow. The bond forms when the adhesive is soft enough to flow (i.e. it is wet) and it has strength when it is hard enough to resist flow when under an applied stress. This is due to van der Waals forces, which determine the bond's strength. PSAs are either permanent or removable, and are used for labels, tape, damping films, 'blu-tack', plastic wrap ('cling film') and plasters.

The strength of adhesion depends on many factors, including the means by which adhesion occurs. Adhesion can occur mechanically by the adhesive running into the pores of the surface, or by chemical mechanisms. A chemical bond may occur between the adhesive and surface. Electrostatic or van der Waals forces may hold the surfaces together. Adhesion may also be due to the moisture-aided diffusion of the adhesive into the surface followed by hardening. Some strong adhesives are important in modern construction and industry.

**Medical Applications:** Adhesives are used in various medical applications including prosthetic adhesives for catheters, dentures and cosmetic purposes. PSAs are often used in skin contact uses, such as wound dressings, ECG electrodes, and analgesic and transdermal drug patches. LCMs may also be used in medical equipment due to its strong bonding and ability to withstand high temperatures.

**Related Article:** Gel

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**Adiabatic RF pulse**

(Magnetic Resonance) The adiabatic phenomenon in magnetic resonance was described as early as 1946 by Bloch in experiments of nuclear induction (Bloch, 1946; Bloch et al., 1946).

In the common amplitude-modulated RF pulses used widely in magnetic resonance, the carrier frequency of the applied RF field is held constant, while the RF amplitude is modulated. In this way, the resulting flip angle of the net magnetisation becomes dependent on the pulse duration and the $B_1$ amplitude. In contrast, adiabatic RF pulses modulate both the frequency and amplitude of the $B_1$ field (Tannus and Garwood, 1997).

When a RF field is applied, a spin population in the rotating frame experiences an effective field $B_{\text{eff}}$, which is composed of the $B_1$-field vector, and a vector in the $B_0$ direction with amplitude proportional to the RF frequency offset. The effective field can change direction (rotate) in space if the applied RF field is frequency- and amplitude modulated (Figure A.22).

If the effective magnetic field vector rotates sufficiently slowly during the radiofrequency pulse, the net magnetisation vector will precess around the effective magnetic field vector and thus follow it.
ARSC was established by the Medicines (Administration of Radioactive Substances) Regulations 1978 and currently consists of 21 members, the majority of which are medical doctors. Medical physicists, radiopharmacists, radiographers and other staff groups also sit on the committee. The committee can also recommend the revoking or suspension of a previously issued licence.

Hyperlinks: ARSC website: www.arsac.org.uk

ADP (adenosine diphosphate)
(Magnetic Resonance) See Adenosine diphosphate (ADP)

Adverse effects
(Nuclear Medicine) Adverse effects, also known as side effects, refer to unwanted negative effects following diagnostic treatment or therapy. Radiopharmaceuticals may have an affinity for organs or biochemical processes other than the intended target organ/process. Organs, other than target organs, that accumulate the administered radiopharmaceutical are referred to as critical organs. The amount of administered activity is limited by the dose to the critical organs. Following an administration three doses are of particular interest:

1. Total body dose, which is correlated to the risk of leukaemia and cancer
2. Gonadal dose as a measure of the hereditary effects
3. Dose to critical organs which can be several times larger than the total body dose. Organs, and their respective tissues, vary in their sensitivity to radiation and a special consideration must be given to the critical organs.

Special consideration should always be taken for pregnant women. The effects of fetus irradiation depends strongly on how far the pregnancy has progressed and the effects may manifest as mental retardation, smaller head size and an increased risk of carcinogenesis.


Adverse effects
(Radiation Protection) Ionising radiation causes biological damage at a cellular level. When such damage is expressed as symptoms either in the person exposed that has not been planned (i.e. not associated with radiotherapy), or is seen in the exposed person’s progeny, it may be referred to as adverse effects, or adverse radiation effects.

For more details, see Bioeffects.

Related Articles: Bioeffects, Adverse radiation effects

Adverse effects
(Radiotherapy) Since any radiation treatment inevitably also affects normal tissue, radiotherapy may cause radiation-induced complications which are known as adverse or side effects. These effects depend on the general status of the patient [e.g. age, co-morbidities, performance status (Karnovsky index)], the clinical situation (tumour site, type, stage, classification, treatment scheme, combined treatment modalities, previous treatments) and on the physical and technical treatment parameters such as radiation type, beam quality, target volume and in particular the spatial and temporal dose distribution.

Adverse effects may occur during the treatment (acute effects, typically up to about 90 days after start of treatment) or months and years after completing the course of treatment (late effects). Adverse effects range from mild effects like tiredness, mood swings, mild forms of nausea, light skin reactions, up to more severe effects such as skin ulceration, and the potential induction of cardiovascular diseases and secondary cancer (carcinogenesis). Typical clinical
examples of acute effects are radiation-induced dermatitis, mucositis, and bone marrow depletion. Clinical late effects include telangiectasia, fibrotic reactions (skin, lung), and in rare cases when exceeding tolerance doses, osteopathy and radiation myelitis.

Acute effects occur often in rapidly proliferating tissue (e.g. mucosa) where radiation deteriorates the balance of cell production in the germinative tissue layers and the cell loss following the final differentiation into functional cells. The regulation mechanism of this balance is not yet fully understood. The stem cell pool itself seems to trigger the proliferation.

Three different categories of interacting effects seem to be responsible for the late response of tissue: (1) in the target cell model the manifestation of tissue damage depends on the characteristics of individual target cells (i.e. proliferation kinetics, repair capacity) and tissue structure; (2) another mechanism is the indirect or reactive effect, e.g. damage of vasculature may be followed by destruction of the parenchyma cells of an organ; (3) finally, radiation may interact at a molecular level via specific signalling pathways and activation of gene expression to induce growth factors and certain proteases. The combination of all three mechanisms leads to the manifestation of tissue radiation response. In general, late effects may occur in tissues with low proliferation rates like the connective tissue and brain.

In the early days of radiotherapy when conventional x-ray and Cobalt units were used, severe side effects, in particular skin reactions were observed. Holthusen in his pioneering work (1936) on radiotherapy optimisation established the sigmoidal dose–response curves for tumour and normal tissue which defines the general strategy of radiotherapy; balancing tumour cure against the probability of adverse effects. Hence, minimising the adverse effects whilst achieving high tumour cure probabilities calls for precisely conforming the dose distribution to the shape of the target volume.

Nowadays, with advanced technologies such as, 3D-CRT and IMRT based on image-guided individual treatment planning and delivery the occurrence of adverse effects is rare. Consideration of the potential induction of adverse effects is an essential part of the radiotherapy treatment process, particularly at the planning stage where the doses to all organs at risk are evaluated alongside that given to the tumour. This has been aided by the availability of dose-volume histograms (DVH) in modern computerised treatment planning systems. Many centres have correlated the incidence of late adverse events with DVH parameters to obtain ‘dose objectives’ for organs at risk to use as a guide for the planning process. All clinical trials with a radiotherapy component now include some guidance on doses to organs at risk in their protocols.

The significance and probability of acute adverse effects can hardly be predicted in the individual case, but mostly they disappear shortly after the course of treatment. Initial information by the physician, appropriate behaviour and sometimes supportive medication may help the patient to cope with the acute adverse effects.

**Abbreviations:**
- 3D-CRT = Three-dimensional conformal radiotherapy
- IMRT = Intensity modulated radiation treatment

**Related Articles:** Probability of complications, Sigmoid dose–response curve


**Adverse radiation effects**
*(Radiation Protection) See Adverse effects*

**AEC (Atomic Energy Commission)**
*(General) See Atomic Energy Commission (AEC)*

**AEC (automatic exposure control)**
*(Diagnostic Radiology) See Automatic exposure control (AEC)*

**Affinity**
*(Nuclear Medicine) A measure of the strength of binding of a ligand to another molecule. The reciprocal of affinity is called $K_d$ (the equilibrium dissociation constant). Thus the higher the affinity that the ligand has for the receptor, the lower its $K_d$ will be.

**Afterglow**
*(Radiation Protection) The afterglow phenomenon occurs in luminescent materials, e.g. scintillators. The energy of ionising radiation (x-ray or gamma photons, charged particles) absorbed in the scintillator causes excitation of orbital electron energy states in the material. The average lifetime of these excited energy states tends to be $\sim 10^{-8}$ s, before decay of the electrons back to the ground state occurs involving the emission of optical radiation (scintillation) in the visible range. This process is called fluorescence. Sometimes the excited state is metastable and its average lifetime is longer, e.g. from microseconds to hours, depending on the scintillator material. The de-excitation is delayed in such materials and this process is called afterglow or phosphorescence. Afterglow is a source of background light (noise) in scintillation detectors.*

**Related Articles:** Scintillator, Scintillation detector


**Afterloading**
*(Radiotherapy, Brachytherapy)*

**Source Handling and Loading:** Brachytherapy source/s must be handled and loaded into the applicators for treatment and, over time, many methods have been used. These methods have been developed primarily to reduce the dose to the personnel, but also to improve the quality of the treatment itself. Afterloading of brachytherapy sources is performed in the following steps:

1. Applicators, needles, catheters, etc. are inserted
2. Correct applicator positions are verified using dummy sources
3. The sources are inserted into the applicators

Generally, afterloading techniques make it possible to devote time to correct placement of applicators and verification of applicator positions.

**Related Articles:** Brachytherapy, Source loading in brachytherapy, Manual loading, Manual afterloading, Remote afterloading, Remote afterloading unit

**Air**
*(General)*

<table>
<thead>
<tr>
<th>Property</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Molar mass</td>
<td>0.029 kg/mol</td>
</tr>
<tr>
<td>Density at STP</td>
<td>$\sim 1.2$ kg/m$^3$</td>
</tr>
<tr>
<td>Melting point</td>
<td>$\sim 63$ K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>$\sim 77$ K</td>
</tr>
<tr>
<td>CT number</td>
<td>$\sim 1000$ HU</td>
</tr>
</tbody>
</table>
Air is a mixture of gases making up the Earth’s atmosphere. Dry air consists of around 78.08% nitrogen, 20.95% oxygen, 0.93% argon, 0.038% carbon dioxide, and trace amounts of other gases. Air also contains a variable amount of water vapour, approximately 1%. Unfiltered air additionally includes particulates and industrial pollutants. The atmosphere acts to protect life by retaining heat via the greenhouse effect, absorbing ultraviolet solar radiation and minimising temperature extremes. The atmosphere gradually decreases in density with distance from the Earth’s surface, and it is normally categorised into several layers. The troposphere is the lowest layer of the atmosphere extending 7–17 km from the surface and containing around 80% of the total mass of the atmosphere. The average temperature and pressure of air at sea level is 288 K and 101.3 kPa, respectively.

Related Articles: Air equivalent composition, Air gap, Air kerma, Air kerma strength, Air-cored transformer, CT number, Entrance surface air kerma (ESAK), Equivalent tissue air ratio (ETAR), Free air ionisation chamber, In air calibration factor, Mean absorbed dose to air, Reference air kerma rate – RAKR, SAR (scatter air ratio), TAR (tissue air ratio)

Air equivalent composition

(General) A material with an air equivalent composition is designed to sufficiently mimic the chemical and physical properties of air depending on the purpose it is designed for.

Medical Applications: In medical physics, air equivalent materials can be used to construct the walls and electrodes of ionisation chambers, such that the number of ionisation events is similar to that generated in a free-air ionisation chamber. Air equivalence requires both the mean mass energy absorption coefficient of the photon spectrum present and the mean mass collision stopping powers of the secondary electron spectrum present to be the same as that of air. Certain plastics are found to be suitable with a graphite surface layer to enable electrical conduction.

Related Articles: Air, Free-air ionisation chamber, Plastic

Air gap

(Diagnostic Radiology) The air gap is space or distance added between a patient’s body and the x-ray image receptor as shown on Figure A.23.

The air gap reduces the intensity of the scattered radiation reaching the receptor. This is not by filtration or absorption but a geometric effect in which the scatter diverges at a greater rate than the primary x-ray beam. Therefore, the intensity of the scattered radiation is reduced when it reaches the receptor.

Air gaps are present when geometric magnification techniques are used, as in magnification mammography. Although magnification is used to enhance image detail, there is the added benefit of reduced scatter reaching the receptor.

Air kerma

(Radiation Protection) Kerma (kinetic energy released per unit mass) is used to describe energy loss in a medium. Thus air kerma represents the kinetic energy transferred to charged particles per unit mass of irradiated air when indirectly ionising (uncharged) radiations such as photons traverse the volume of air.

For more details see the article on Kerma

Related Article: Kerma

Air kerma

(Radiotherapy) The kerma (K) is defined as the mean energy transferred from the indirectly ionising radiation to charged particles in the medium per unit of mass at a point of interest without concern as to what happens after this transfer. As the energy transferred to charged particles depends on the irradiated medium, a statement of kerma is incomplete without a reference to the material concerned. Air kerma results from photon interactions with air.

It is worthy to show a relationship of air kerma with exposure, another dosimetric quantity related with air interactions.

Air kerma is related to the energy fluence by

\[ K_{air} = \psi \mu_{en} / \rho \]

where

- \( \psi \) is the fluence energy
- \( \mu_{en} / \rho \) is the mass energy transfer coefficient

and the exposure is given by

\[ X = \psi \left( \frac{\mu_{en}}{\rho} \right)_{air} \frac{e}{W_{air}} \]

where

- \( \psi \) is the fluence energy
- \( (\mu_{en}/\rho)_{air} \) is the mass energy absorption coefficient of air
- \( W_{air} \) is the average energy required to produce an ion pair in air
- \( e \) is the electron charge
Air kerma strength

\[ K_{\text{air}} = \frac{XW_{\text{air}}}{e} \left( \frac{\mu_{\text{air}}}{\mu_{\text{H}}^0} \right) = \frac{XW_{\text{air}}}{e} \left( 1 - g \right) \]

where \( g \) is the fraction of electron energy lost in bremsstrahlung production.

The fraction \( g \) is significant only at high energies. Its value for \( ^{60}\text{Co} \) photons is 0.003.

**Related Articles:** Kerma, Collision kerma

**Air kerma strength**  
(*Radiotherapy, Brachytherapy*) Calibration of source strength is a very important part of a comprehensive brachytherapy quality system. The instruments, ion-chambers and electrometers, used for source strength determinations, should have calibrations that are traceable to national and international standards.

**Specification of Source Strength for Photon Emitting Sources:** Source strength for a photon emitting source can be given as a quantity describing the radioactivity contained in the source or as a quantity describing the output of the source:

1. Specification of contained activity
   a. Mass of radium; mg Ra
   b. Contained activity; Ci, Bq
2. Specification of output
   a. Equivalent mass of radium; mg Ra eq
   b. Apparent activity
   c. Reference exposure rate
d. Reference air kerma rate
e. Air kerma strength

Task Group 43 of the AAPM defines ‘air-kerma strength’:

‘Air-kerma strength has units of \( \mu\text{Gy m}^2/\text{h} \) and is numerically identical to the quantity Reference Air Kerma Rate recommended by ICRU 38 and ICRU 60’ (ICRU 38, 1985; ICRU 60, 1998). For convenience these unit combinations are denoted by the symbol \( U \) where \( U = 1 \mu\text{Gy m}^2/\text{h} = 1 \text{cGy m}^2/\text{h} \). The National Institute of Science and Technology (NIST) maintains the US primary air-kerma standards for x-rays in the energy range of 10–300 keV and for photon-emitting radionuclides such as \(^{137}\text{Cs}, \; ^{192}\text{Ir}, \; ^{103}\text{Pd} \) and \(^{125}\text{I} \). Air-kerma strength, \( S_K \), is the air-kerma rate, \( K_\delta(d) \), in vacuo due to photons of energy greater than \( \delta \), at a distance \( d \), multiplied by the square of this distance, \( d^2 \). The distance \( d \) is measured along a perpendicular bisector of the source.

\[ S_K = K_\delta(d) \cdot d^2 \]

In modern brachytherapy dosimetry, reference air kerma rate or air kerma strength is the quantity used to calculate absorbed dose.

See **Source strength** for a full description of specification of source strength.

**Abbreviations:** AAPM = American Association of Physicists in Medicine and ICRU = International Commission on Radiation Units and Measurements.

**Related Articles:** Source strength, Mass of radium, Contained activity, Equivalent mass of radium, Apparent activity, Reference air kerma rate (RAKR)


**Air-correlated transformer**  
(*Diagnostic Radiology*) See **Transformer**

**AIUM**  
(*Ultrasound*) The American Institute for Ultrasound in Medicine was formed in the early 1950s to promote the emerging science of medical ultrasound and its diagnostic and therapeutic applications. Since then it has developed into a leading multidisciplinary organisation with a membership of physicians, technologists, sonographers, scientists, physicists and engineers. The AIUM works with users, manufacturers and government to produce guidelines for the safe and effective use and practice of medical ultrasound, many of which are available through its website. The Institute produces the *Journal of Ultrasound in Medicine* and runs a large annual meeting and other smaller educational meetings in North America.

**Hyperlink:** AIUM: www.aium.org

**ALARA**  
(*Ultrasound*) The principle of ALARA (as low as reasonably achievable), as applied in ultrasound, includes using low output powers where patient safety is an issue and using lower intensity modes (e.g. B-mode) before adding higher output modes such as colour flow and spectral Doppler imaging.

**ALARA**  
(*Radiation Protection*) See **As low as reasonably achievable**

**ALARP**  
(*Radiation Protection*) See **As low as reasonably practicable**

**Algorithm**  
(*General*) An algorithm is a series of well-defined mathematical operations, i.e. a calculation method. Each operation or algorithm state yields a result that may affect the choice of the subsequent stages, as in a flow chart. A simple flow chart for a scintillation camera reconstruction is seen in **Figure A.24**.

**Aliasing**  
(*Magnetic Resonance*) Aliasing, ‘phase wrap’ or ‘foldover’ is an image artefact that occurs in MRI where there is anatomy outside

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**Figure A.24** A flowchart describing the reconstruction process of raw data acquired by a scintillation camera.
Application of a phase encoding gradient creates a spatial variation in temporal frequency given by

$$\Delta \omega = 2\pi G_y \gamma$$  \hspace{1cm} (A.12)

where
- $y$ is distance from isocentre along the direction of the applied gradient $G_y$.
- $\gamma$ is the gyromagnetic ratio.

On completion of phase encode step $i$ of duration $T$, the variation of phase with location $y$ in the phase encoding direction is

$$\phi_i(y) = 2\pi G_y T \gamma$$  \hspace{1cm} (A.13)

If $G_{max}$ is the magnitude of the maximum gradient applied, the maximal variation of phase with position is

$$\phi_{max}(y) = 2\pi G_{max} T \gamma$$  \hspace{1cm} (A.14)

On completion of the maximal phase encoding step, the phase difference between two adjacent pixels separated by the pixel width $\Delta y$ is 180°:

$$\Delta \phi_{max}(\Delta y) = \pi = 2\pi \delta y \gamma G_{max}$$  \hspace{1cm} (A.15)

After rearrangement,

$$[G_{PhaseMax}] = \frac{1}{2\Delta y \cdot T \gamma}$$  \hspace{1cm} (A.16)

Over the course of the scan, the phase encoding gradient ranges between $-G_{PhaseMax}$ to $+G_{PhaseMax}$. For $N_p$ pixels in the phase encoding direction, the phase encoding gradient changes by

$$\Delta G_{phase} = \frac{2G_{PhaseMax}}{N_p} = \frac{2}{\Delta y \cdot T \gamma N_p}$$  \hspace{1cm} (A.17)

at each phase encode. The change in phase at any given location $y$ between phase encodings is then (combining [A.12] and [A.16])

$$\Delta \phi(y) = \frac{2\pi y}{\Delta y \cdot N_p}$$  \hspace{1cm} (A.18)

As

$$N_p = \frac{\text{FoV}}{\Delta y}$$  \hspace{1cm} (A.19)

The phase change between phase encodings at any location $y$ can be expressed as

$$\Delta \phi(y) = \frac{2\pi y}{\text{FoV}}$$  \hspace{1cm} (A.20)

For anatomy outside of the field of view in the phase encoding direction,

$$|y| > \frac{\text{FoV}}{2}$$  \hspace{1cm} (A.21)

so the phase change between phase encodings (A.19) is $> \pi$. Phase changes with magnitude $> \pi$ are ambiguous, so phase encoding of points at all locations

$$y = n \cdot \frac{\text{FoV}}{2} \quad n = 0, 1, 2, \ldots$$  \hspace{1cm} (A.22)

is ambiguous and these points are mapped back into the FoV causing the foldover artefact.

Similarly in the frequency encoding direction, any anatomy beyond the edges of the user set FOV generates signal at frequencies that violate Nyquist sampling requirements. However, aliasing in the frequency encoding direction can be prevented simply by low pass temporal filtering to exclude signal originating from beyond the FOV.

**Aliasing (Ultrasound)** Aliasing is a well-known phenomenon to all that has studied sampling theory. The Nyquist theorem states that the sampling frequency must be at least twice the highest frequency component of the input signal, or else aliasing will occur. By this, it is meant that the reconstructed signal will appear at a lower frequency than it originally had. Usually this is illustrated with two sinus-signals, where the slower one is reconstructed from sampling of the faster waveform at the equidistant points marked in the figure.

Usually in connection with Doppler measurements, this illustration is not entirely applicable. In this limiting case, when the input frequency is exactly half the sampling frequency, the same frequency will be reconstructed. If the input frequency $f_0$ increases an amount $\epsilon$, the reconstructed frequency will be $f_0 - \epsilon$. If that would be the case in Figure A.26, illustrating carotid flow, the peak velocities would appear as ‘folded’ and interfere with the waveform. Instead they show up as negative velocities. Why it appears different in a Doppler sonogram is due to the fact that usually a quadrature detection is made. This results in a pair of signals (I and Q channels), and the subsequent Fourier transformation can be thought of as performed on a complex signal with these as the real and imaginary parts, respectively. The result is a nonsymmetric spectrum, with a representation of both negative and positive velocities, as opposed to an FFT performed on a real signal, which results in a symmetric spectrum. The image of the spectrum of a sampled signal as one that repeats itself around multiples of the sampling frequency is probably more useful in this case. Aliasing can also be found in the same way in colour Doppler images, see Figure A.27.
gamma-ray, x-ray, or low energy beta-ray, and ions such as carbon or helium. The typical continuously bending shape of cell survival curves has been successfully interpreted based on radiation dose. The initial radiobiology experiments established the relation of cell survival to radiation dose. For example, Curtis (1986) in his unified repair model of cell killing proposed two different causes that are ultimately responsible for cell kill: the reparable and the nonreparable lesions. The nonreparable lesions such as DSB, or in terms of the target theory the single-hit lethal effects, are associated with the linear term \( \exp(-\alpha D) \) of cell killing, whereas the repair itself is a balance of successfully repaired events and binary misrepair, the latter giving rise to the \( \exp(-\beta D^2) \) term. However, all these mechanistic approaches to understand the cell survival became to some extent obsolete, particularly in the light of molecular radiobiology.

In the clinical environment the alpha/beta-ratio is widely used to assess the importance of acute and late effects in normal tissue. High alpha/beta-values (usually 10), i.e. nearly straight survival curves, represent tissues with limited potential for repair from radiation damage. On the other hand, normal tissues with low alpha/beta-values (usually 3), i.e. with a significant shoulder of the survival curve at low doses, are characterised by their high recovery potential. Hence, tissues with low alpha/beta-ratios are sensitive to the dose fractionation scheme (fractionation effect) whereas those with high alpha/beta-ratios are hardly affected by dose fractionation. Furthermore, the alpha/beta-ratio is clinically applied to calculate isoeffective doses when changing the fractionation schedule. For example, replacing a fraction dose \( d_1 \) by \( d_2 \), the total dose for the new fractionation scheme is

\[
D_2 = D_1 \frac{(\alpha/\beta + d_1)}{(\alpha/\beta + d_2)}
\]

As many normal tissue tolerance values refer to a standard fraction dose of 2 Gy the formula can be used to assess the radiation response when applying an alternate treatment regime.

**Abbreviations:** DSB = Double strand break and SF = Surviving fraction

**Related Articles:** Linear-quadratic model, Radiobiological models, Adverse radiation effects, Biological effective dose, Surviving fraction

Alpha emission
(Nuclear Medicine) In α-decay an α-particle, consisting of two neutrons and two protons, is emitted from a nucleus. The α-particle has high kinetic energy, typically 4–8 MeV, but due to frequent interactions with the surrounding material the range is limited to as little as a few μm in solid materials. Therefore most α-emitters are never used for clinical imaging. On the other hand, α-particles are very efficient in inducing cell death and are therefore an option when designing therapeutic methods.

Alpha particle emitter
(Radiation Protection) An alpha particle emitter is any element that emits alpha particles as a result of radioactive decay. Alpha emitters either occur naturally — there are three series of naturally occurring substances (uranium, thorium and actinium) – or can be created by bombarding specific elements with high-energy particles or by fusion induced by a neutron in a nuclear reactor (or nuclear device).

Related Articles: Alpha particles, Nuclear fusion

Alpha particles
(General) Alpha particles are charge particles, with a charge of two and a mass of four, which consist of two protons and two neutrons bound together. Alpha particles are identical to the nuclei of helium atoms (\(^4\)He), and either the symbol α (first letter of Greek alphabet) or \(\text{He}^{2+}\) can be used to denote an alpha particle.

Alpha particles were discovered by Ernest Rutherford (1871–1937) in 1899 and are highly ionising due to their mass and double charge.

- Mass 6.644656 × 10^{-27}, equivalent to 3.727738 GeV.
- Alpha particles are emitted in some forms of radioactive decay (alpha decay) where the radionuclide has an excess of neutrons and protons. Beams of alpha particles can also be produced by ionising helium atoms to produce alpha particles and accelerating them in a cyclotron.

Related Articles: Alpha particle emitter, Radioactivity, Radioactive decay.

Alpha radiation
(General) Alpha radiation is composed of alpha particles, emitted from radionuclides undergoing radioactive decay though the decay process normally referred to as ‘alpha decay’. Alpha radiation can also be generated by ionising helium atoms to produce alpha particles and accelerating them in a cyclotron.

Alpha particles are charged particles, with a charge of two and a mass of four, which consist of two protons and two neutrons bound together. Alpha particles are identical to the nuclei of helium atoms (\(^4\)He), and either the symbol α (first letter of Greek alphabet) or \(\text{He}^{2+}\) can be used to denote an alpha particle.

Related Articles: Alpha decay, Alpha particles, Radioactive decay, Radionuclide.

Alternating current (AC)
(General) Alternating current is a form of electric current in which the direction of flow changes or alternates regularly. This distinguishes it from ‘direct current (DC)’, where the current flows in one direction and is typically constant in value.

Alternating current is easy to generate, and AC power may be transferred efficiently using transformers. Most domestic electrical power is AC power, typically supplied at a potential of 110–230 V RMS.

Alternating current sources are usually sinusoidal in form and produced at a frequency of 50–60 Hz (Figure A.29), though the term AC and the theory of AC conduction apply to currents and voltages at any frequency.

It is necessary to define AC voltage and current values in terms of their DC equivalents, so that the power provided into a resistive load will be the same. This is defined as the RMS or ‘root-mean-square’ AC value, and for sinusoidal signals is \(1/\sqrt{2}\) of their peak values.

The electrical power (in watts) in a DC circuit can be calculated simply by multiplying the potential difference across a load (in volts) by the current through the load (in amps). Similarly, the apparent power in an AC circuit can be deduced using the \(V_{\text{rms}}\) and \(I_{\text{rms}}\) values.

However, true AC electrical power calculations are more complex as the AC current does not necessarily flow ‘in phase’ with the applied potential.

The relative phases of the AC potential and current must also be taken into account where a load possesses reactive properties (capacitance or inductance). In such cases the power delivered to the load is given by

\[
\text{AC power} = \text{AC voltage}_{\text{rms}} \times \text{AC current}_{\text{rms}} \times \cos(\text{phase angle})
\]

where the phase angle represents the difference in phase between the voltage and current.

Related Articles: Alternating voltage, Apparent power, Direct current, Direct voltage

Alternating voltage
(General) A form of electrical potential which changes polarity regularly, and is usually sinusoidal in amplitude. This distinguishes it from ‘a direct voltage’ where the potential remains at a constant fixed value.

Alternating voltages cause alternating currents (AC). Most domestic electrical power is AC power, typically supplied at a potential of 110–230 V RMS with a frequency of 50–60 Hz.

Alternating voltages may be specified in terms of their peak-to-peak voltage swing (\(V_{\text{pk-pk}}\)), the peak swing (\(V_{\text{pk}}\), or by the rms or root-mean-square voltage (\(V_{\text{rms}}\)) which has the equivalent power capability as the DC voltage of the same value.

For a sinusoidal alternating voltage: \(V_{\text{pk-pk}} = 2 \times V_{\text{pk}} = 2 \times \sqrt{2} \times V_{\text{rms}}\)

Abbreviations: AC = Alternating current, DC = Direct current and RMS = Root-mean-square.

Related Articles: Alternating current, Apparent power, Direct current, Direct voltage
Alum in film processing

(Diagnostic Radiology) Alum is aluminium salt. Such salt is used in x-ray film processing – especially as hardener (e.g. potassium alum). The aluminium salt prevents excessive softening of the emulsion (which may damage it during the washing or drying processes).

Related Articles: Hardening agent, Developer

Aluminium

(General)

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Al</th>
</tr>
</thead>
<tbody>
<tr>
<td>Element category</td>
<td>Metal</td>
</tr>
<tr>
<td>Mass number A</td>
<td>27</td>
</tr>
<tr>
<td>Atomic number Z</td>
<td>13</td>
</tr>
<tr>
<td>Atomic weight</td>
<td>26.9815 g/mol</td>
</tr>
<tr>
<td>Electronic configuration</td>
<td>1s^2 2s^2 2p^6 3s^2 3p^1</td>
</tr>
<tr>
<td>Melting point</td>
<td>933.47K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>2792 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>2.70 g/cm^3</td>
</tr>
</tbody>
</table>

History: Aluminium salts were used in Ancient Greek and Roman medicine for wound dressing. Aluminium was first extracted in pure metal form by Friedrich Wöhler in 1827, by reacting potassium and anhydrous aluminium chloride.

Although Aluminium is the official international spelling, Aluminium is widely used in the United States and Canada, and is accepted as an alternative spelling by many chemical societies.

Isotopes of Aluminium: Pure aluminium is highly reactive and is therefore uncommon in nature. However, it is abundant in compound with other molecules, particularly as oxides and silicates. Stable^{27}Al is by far the most common naturally occurring isotope, with more than 99.9% relative abundance. The remainder consists of radioactive^{26}Al, produced in the atmosphere by cosmic ray photons reacting with argon. There are seven further isotopes, with mass numbers between 23 and 30, but these do not occur naturally and must be synthesised.

Medical Applications: Solid-state lasers – Aluminium forms part of the yttrium-aluminium-garnet (YAG) crystal that is used in many solid-state medical lasers. This crystal can be doped with neodymium (Nd-YAG), erbium (Er-YAG) or other rare earth elements, to produce light of different wavelengths. Nd-YAG and Er-YAG lasers (with wavelengths of 1064 and 2940 nm, respectively) have many uses in medicine, including ophthalmology, dentistry and cosmetic corrective treatments.

Radiation shielding: Aluminium is traditionally used to attenuate ionising radiation (particularly beta radiation), both for radiation protection purposes and in the shielding of equipment from external radiation that may affect its performance. The shielding performance of any material can be expressed in terms of ‘mm Al’, the thickness of aluminium that would be required to provide the equivalent reduction in radiation flux.

Aluminium equivalent

(Diagnostic Radiology) Both the inherent and total filtration through which an x-ray beam passes can be expressed in mm of aluminium. This, the aluminium equivalent, is the thickness of aluminium that would provide the same absorption/filtration.

Related Articles: Filtration inherent, Filtration total

AMBER

(Diagnostic Radiology) Advanced multiple-beam equalisation radiography (AMBER) is a technique developed by Kodak to reduce the problem of the wide range of exposures coming from a patient’s body, especially for thoracic imaging. In the chest the low density lung areas produce high exposures to the image receptor and the more dense mediastinum produces low exposures. The problem is that this range of exposure can exceed the optimum contrast producing range, or latitude, of radiographic film.

The AMBER system uses a scanning slit x-ray beam which is divided into 21 beam segments. The beam intensity in each segment is modulated based on the intensity of the beam penetrating the patient’s body in that area and measured by individual x-ray detectors. The output from each detector controls the corresponding modulator. The AMBER principle and some modulator settings for different anatomical areas are shown in Figure A.30.

For areas of high intensity, such as in the lungs, the beam modulator attenuates more of the beam and reduces the intensity reaching the receptor. The result is a reduced range of exposures reaching the receptor from the different areas of the chest.

Ambient lighting

(Diagnostic Radiology) The level of ambient lighting or illumination in a room can have an effect on the visibility of contrast in displayed digital images. Some display devices or monitors have photocells which measure the light in the room and change the brightness of the monitor to produce optimum viewing conditions. AAPM reports some typical ambient lighting levels for a number of image display devices (e.g. typical 2–10 lux for x-ray diagnostic reading stations).

Americium
(General)

<table>
<thead>
<tr>
<th>Element category</th>
<th>Actinide</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mass number $A$</td>
<td>241 and 243 (no stable isotope known)</td>
</tr>
<tr>
<td>Atomic number $Z$</td>
<td>95</td>
</tr>
<tr>
<td>Atomic weight</td>
<td>241 and 243</td>
</tr>
<tr>
<td>Electronic configuration</td>
<td>$1s^2 2s^2 2p^6 3s^2 3p^6 3d^{10} 4s^2 4p^6 5s^2 4d^{10}$</td>
</tr>
<tr>
<td>Melting point</td>
<td>1449 K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>2880 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>12000 kg/m³ (12 g/cm³)</td>
</tr>
</tbody>
</table>

**History:** Americium is an artificial metal that was first obtained by Glenn Seaborg and colleagues in 1944. The team, who were based at the University of Chicago, created the isotope $^{241}$Am by bombarding plutonium with neutrons. Named for the Americas, Americium is now used worldwide in smoke detectors and as a neutron source in industrial moisture gauges.

**Isotopes of Americium:** All of americium’s isotopes ($^{237}$Am through to $^{246}$Am) are radioactive. The isotope of interest in medicine is $^{241}$Am.

<table>
<thead>
<tr>
<th>Isotope of Americium</th>
<th>$^{241}$Am</th>
</tr>
</thead>
<tbody>
<tr>
<td>Half-life</td>
<td>432.2 years</td>
</tr>
<tr>
<td>Mode of decay</td>
<td>$\alpha$, plus $\gamma$ in $\approx 80%$ of decays</td>
</tr>
<tr>
<td>Maximum decay energy, $E_{\text{max}}$</td>
<td>$\alpha$: 5.638 MeV, $\gamma$: 13.9 keV or 59.5 keV</td>
</tr>
</tbody>
</table>

The decay product, neptunium-237, is also radioactive with a two million year half-life.

**Medical Applications:** Fluorescence thyroid scanning – external sources of $^{241}$Am can be used to bombard stable iodine in the thyroid with gamma rays, causing it to fluoresce. The fluorescence photons released quantitatively correlate with the iodine content of the imaged tissue, such that a plot of the thyroid’s iodine content can be produced (low iodine content can be an indicator of thyroid cancer). As this technique involves minimal radiation exposure, it may prove particularly useful in the diagnosis of paediatric/pregnant patients, but unfortunately it is not widely available as yet.

Gynaecological Brachytherapy – In the past $^{241}$Am intracavitary applicators were used for the treatment of gynaecological cancer, but now $^{192}$Ir and $^{137}$Cs applicators are more usual (their shorter half-lives allow for easier disposal and reduced security risk).

**Related Articles:** Fluorescence, Brachytherapy, Brachytherapy sources

**A-mode**
(Ultrasound) The amplitude mode (A-mode) is the most basic form to display ultrasound pulse-echo measurements. The echoes detected by a transducer in a single line measurement are displayed on an oscilloscope with the echo-amplitudes on the y-axis and the time delay (depth) on the x-axis (Figure A.31).

A-mode measurements were common on early ultrasound systems. Nowadays the technique is used for detection of sinus infection and when measuring eyeball length and skin thickness. In some systems, the operator can select the speed of sound appropriate to the tissue under investigation to optimise distance measurement accuracy.

**Related Articles:** B-mode, M-mode

**Amorphous selenium**
(Diagnostic Radiology) Selenium (Se) is a nonmetal chemical element with an atomic number ($Z$) of 34. It is found in rare minerals such as crooksite and clausenthalite but is generally obtained as a by-product from the anode metal in electrolytic copper refineries. It exhibits both photoconductive and photovoltaic properties; in the former the number of charge carriers increases when it absorbs certain wavelengths of electromagnetic radiation increasing the electrical conductivity, in the latter incident light generates a voltage across the material.

In medical imaging selenium is used as a photoconductor in flat-panel detectors as it is easily manufactured into a large area continuous photoconductor layer with low dark current and high x-ray sensitivity. It is the most widely employed photoconductor in flat-panel direct-conversion x-ray imaging. In other industries selenium has been widely used as a photoconductor in photocopiers, in a traditional x-ray imaging technique called xeroradiography, and as a photovoltaic material in solar cells. Although selenium is a very useful photoconductor it has many other applications including, converting alternating current to direct current in a rectifier and as a decolourising green glass and producing ruby coloured glass.

The structure of amorphous selenium (a-Se) is described by the random chain model as a twofold coordinated chain structure in which the angle between two adjacent bonding planes $\phi$ (dihedral angle) is constant in magnitude but randomly changes sign. This creates regions of ring-like and chain-like structures which are randomly distributed throughout the material. This compares to crystalline selenium which occurs in one of two forms, either in rings (As) and doping with chlorine (Cl) in 10–20 ppb range. The arsenic is introduced to prevent the structure from re-crystallising while the Cl compensates for the hole traps introduced by the As. The amorphous nature of selenium is advantageous in the manufacture of flat-panel detectors as it can be easily deposited on a suitable substrate by conventional vacuum deposition techniques to form large area photoconductive film of thicknesses up to 1000 μm. This is in contrast to polycrystalline structures which are difficult to grow large enough to cover large area detectors (e.g. 40 x 40 cm).

**FIGURE A.31** Principle of A-mode display. The display shows the amplitude of echoes and the distance between them. Distance assumes known speed of sound in tissue.
Amorphous selenium photoconductive layer

(Diagnostic Radiology) Selenium (Se) is a non-metal chemical element of atomic number 34. It is found in rare elements such as crooksite and clausthalite and is generally obtained from the anode in electrolytic copper refineries.

Selenium is used as a photoconductor to detect x-rays in flat-panel detectors in medical imaging. It is the most widely employed photoconductor in flat-panel detectors for direct conversion x-ray image detection. The increased number of charge carriers reduces electronic resistance and increases the conductivity.

Figure A.32 shows the electronic band structure of a photoconductor. The valence band immediately below the forbidden energy gap is almost completely full while the conduction band is usually completely empty. When photons of a large enough energy are absorbed by the photoconductive material, a bound electron in the valence band is given enough energy to move to the conduction band across the forbidden region. This creates an electron–hole pair, in which the electron can freely move in the conductive band and the electron ‘hole’ created in the valence band moves through the material similar to that of a physical charged particle. The increased number of charge carriers reduces electronic resistance and increases the conductivity.

Table A.3 lists several properties of stabilised amorphous selenium. X-ray sensitivity of a photoconductor can be defined as the charge collected per unit incident radiation per unit device area. Two factors affect the sensitivity: (1) the radiation energy $W_x$ absorbed by the medium to create a free electron and hole pair; and (2) the x-ray absorption coefficient $\alpha$ of the material. From the absorption coefficient the absorption depth $\delta$ is defined as $1/\alpha t$, which is the depth at which 63% of the incident x-rays are absorbed by the photoconductor. Amorphous selenium is favoured as a photoconductor in flat-panel detectors as it has a high x-ray absorption coefficient due to its high atomic number, better transport properties for electrons and holes compared to other selenium-based compounds and low-dark current due to a large energy gap, $E_g$.

As selenium is used in its amorphous state it can easily be deposited onto the active matrix substrate, used in flat-panel detectors, by conventional vacuum deposition techniques at low temperatures of below 70°C. This is in contrast to crystalline materials, which require advanced and thus expensive manufacturing techniques and high temperatures for both annealing and optimisation, which may thermally damage the substrate below.

Although amorphous selenium is the most widely used photoconductor for flat-panel technology, at present several other alternatives have been investigated. Amorphous silicon (a-Si:H) has a lower atomic number than a-Se ($Z = 14$) which means that the absorption coefficient is much smaller than selenium and consequently a thicker layer would be required for the similar absorption properties making fabrication much more difficult. Polycrystalline photoconductors such as Te and PdI$_2$ have a higher atomic number and have been manufactured in the crystal sizes required. However, their grain boundaries limit charge transportation. Growth of single crystal materials are currently not suitable for flat-panel detectors as they have only been manufactured in diameters of 5–10 cm and are too small for large-area detection.

**Relevant Articles:** Stabilised a-Se, Amorphous selenium, Selenium detector, TFT (thin film technology)


**General**

Ampere

Ampere is the unit for electric current. It is one of the seven SI base units from which all other units can be derived. Consider a setup with two parallel infinitely long conductors separated by 1 m. When a current is applied to the two conductors, they will assert a force on one another. With these settings 1 A
Ampere-second

(Diagnostic Radiology) The ampere-second term is used in diagnostic radiology (mainly as milliampere-second, or mA-s) to describe the charge of the electrons, emitted by the cathode (thermal electrons) and bombarding the target of the x-ray tube:

\[ 1 \text{ A} \cdot 1 \text{ s} = (1 \text{ C/1 s}) \cdot 1 \text{ s} = 1 \text{ C} \text{ (here } 1 \text{ A} = 1 \text{ C/1 s}) \]

This way, the quantity of x-rays is directly proportional to the quantity of charge of the thermal electrons \( Q = \text{mA} \times \text{s} \) – the Anode current (mA) and the time of the exposure (s).

The mA-s is measured as the anode current integrated over the time of the exposure (using an integrator). Usually a 25% change of the mA-s presents an x-ray film image with a clearly visible change in contrast (optical density). Such a change is often related to one exposure point (such points are still used in some of the exposure tables used in the radiography practice).

Related Article: mAs selector

Amplification factor

(Nuclear Medicine) The amplification factor is the signal multiplication factor in a photomultiplier (PM) tube. Two properties determine the amplification factor, namely the number of dynodes and multiplication factor at each dynode. The signal multiplication factor in a PM tube with 10 dynodes and electron multiplication factor of 6 at each dynode is \( \sim 6 \times 10^7 \). It is important that the amplification factor is constant and not dependent on the amount of energy deposited in the photon interaction.

Related Article: Photomultiplier (PM) tubes


Amplifier

(Nuclear Medicine) A nuclear pulse amplifier is used to convert a low-amplitude pulse from a radiation detector to one with sufficient amplitude and the proper pulse shape to drive the pulse-selecting elements of the counting system. The amplifier must have enough gain to drive the pulse selector while still enabling the detector to operate in its most favourable operating range. Its gain must be stable.

The capacitance introduced by a signal cable running from the radiation detector to the amplifier will in many cases significantly attenuate and distort the transmitted electrical impulse. Preamplifiers placed near the detector will minimise this input capacity and may be designed to provide impedance matching at their output so that long cables can be used.

In any modern detectors the signal after the preamplifier is digitised for further processing.


Amplitude attenuation coefficient

(Ultrasound) When sound travels through a medium, its intensity diminishes with distance. Signal amplitude is reduced not only by the spreading of the wave but also by scattering and absorption. The combined effect of scattering and absorption is called attenuation. Ultrasonic amplitude attenuation is the decay rate of the wave as it propagates through material.

The amplitude change of a decaying plane wave can be expressed as

\[ A = A_0 \times \exp(-\alpha z) \]

where

- \( A_0 \) is the unattenuated amplitude of the propagating wave at some location
- \( A \) is the reduced amplitude after the wave has travelled a distance \( z \) from that initial location
- The quantity \( \alpha \) is the attenuation coefficient of the wave travelling in the \( z \)-direction

Analogue image

(General) An image where the darkness varies continuously with the irradiation intensity. An analogue image is also continuous in space, i.e. no pixels. The opposite of the analogue image is the digital image consisting of several image elements, i.e. pixels, and each pixel is associated with a discrete number proportional to its corresponding signal intensity.

In early radiography, films covered with light-sensitive emulsion (silver compound) were used to acquire images. The limiting factor for the spatial resolution in a conventional film is the physical spread of each individual silver compound. The highest possible amount of darkness produced depends on the density of the silver compound and the greyscale varies continuously.

Analogue signal

(General) An analogue signal is a variable signal with continuity in time and amplitude rather than a pulsed or discrete nature. A variation in the signal is a representation of another time varying quantity. In electronics physical properties measured are voltage, phase, frequency and charge. A disadvantage with analogue signalling is that most systems are typically sensitive to noise. When an analogue signal is copied and re-copied or transmitted over long distances the noise increases which eventually renders the signal useless, overshadowed by the noise. A digital signal is much more resistant to noise compared to an analogue signal. Another way to convey an analogue signal is to use modulations of the signal. A signal, typically a sinusoidal carrier wave, has one of its properties modulated, e.g. amplitude or frequency.

Analogue tracer

(Nuclear Medicine) Analogue tracers are compounds created to mimic the properties of natural compounds. An analogue tracer can be tailored so that the tracer only takes part in selected components of a biological process. This can minimise the number of variables in a process, thus increasing the specificity and accuracy of the measurements.

Another use of an analogue tracer is to tailor a compound that it can be labelled with an element (radioisotope) that is not present in the natural biological compound so that the tailored compound can be used to study the behaviour of the natural compound.

The biochemical properties of an analogue tracer are not always identical to the naturally occurring compounds. To compensate for this one uses correlation factors.

One of the most common analogue tracers is FDG (18F-2-fluoro-2-deoxy-D-glucose) which measures glucose metabolism.

Related Article: Isotope effect

Analogue-to-digital converter (ADC)

(General) A problem in conventional pulse height analysis is when some applications require a simultaneous acquisition and separation of events into different voltage or energy windows. In a multi channel analyser (MCA), the pulses are transferred from an analogue signal, which has a continuous range of possible voltage values, to a digital pulse which can be sorted into one of a finite number of energy windows, or channels. This converter circuitry is called an analog-to-digital converter (ADC). If the height of the analogue pulses is in the range of 0–10 V, a 1000 channel analyser would divide the voltage evenly over each channel. For example, channel 1 corresponds to 0–0.01 V, channel 2 to 0.01–0.02 V and so on. For every channel, the MCA has a specific memory storage location, so that each pulse that is registered within a certain energy window is recorded. There are two types of ADC commonly used in nuclear medicine; these are the Wilkinson converter and the successive approximation converter.


Anatomical body planes

(General) To describe anatomical planes imagine a person standing in an upright position and dividing this person with imaginary vertical and horizontal planes. Anatomical planes can be used to describe a body part or an entire body.

Sagittal Plane: Picture a vertical plane that runs through the body from front to back or back to front. This plane divides the body into right and left regions. This plane will pass approximately through the sagittal suture of the skull, and hence, any plane parallel to it is termed a sagittal plane.

Coronal Plane: Think of a vertical plane that runs through the centre of your body from side to side at right angles to the sagittal plane. This plane divides the body into front (anterior) and back (posterior) regions. This plane runs through the central part of the coronal suture or through a line parallel to it; such a plane is known as a coronal plane.

Axial or Transverse Plane: This horizontal plane divides the body into the upper (superior) and lower (inferior) regions by running through the midsection of the body.

Anatomical landmark

(Radiotherapy) Anatomical landmarks refer to distinctive structures in the patient’s anatomy that may be easily localised. Anatomical landmarks are generally classifiable as rigid and mobile. Rigid landmarks are those which do not move as the patient breathes and include bony structures. Mobile landmarks include bronchi which may be used to determine how anatomy moves when the patient breathes.

Treatment Setup: Landmarks for setup for treatment may include

1. External points on the skeleton, which may be used for basic treatment set-up with tattoos and lasers
2. Internal bony structure, which may be used with x-ray imaging for setup
3. Internal soft tissue, which may reveal the extent of motion of soft tissue relative to bony anatomy

Image Registration: Often a multi-modality approach is used in medical imaging, in order to yield extra information compared to a single modality. The images are often registered spatially. This often involves identifying anatomical landmarks in each image, calculating the transformation needed to register them and interpolating/extrapolating for other regions.

Related Articles: Imaging, Multi-modality imaging, Laser localisers

Anatomical noise

(Diagnostic Radiology) Anatomical noise is a psychophysical noise source in radiographic images. It is caused by the projection of overlaying small anatomical structures on the radiograph. These overlaying structures cannot be distinguished by the human observer as specific anatomical detail and can be modelled as another source of image noise as it limits the detection of small pathology.

Figure A.33 illustrates how anatomical and quantum noise affects the detection of pathology by a human observer. It assumes that quantum noise is described by the Rose model and is thus proportional to the absorbed detector dose. As the radiation dose to the detector increases, the quantum noise signal-to-noise ratio increases. However, the reproduction of small anatomical structures also increases in clarity with radiation dose, which in turn increases the anatomical noise.

Although the efficiency and performance of the system can be modelled quantitatively by the modulation transfer function (MTF), noise power spectrum (NPS) and detective quantum efficiency (DQE), the detection of pathology by a human observer relies upon the total noise which they perceive. Consequently, there is much debate over the best way to assess radiological system performance and whether to include qualitative observer-based assessment to allow the inclusion of anatomical noise sources. Such qualitative assessments include receiver operating characteristics (ROC) analysis and visual graded analysis (VGA) and visual graded characteristics (VGC).

Related Articles: Quantum noise, Noise power spectrum (NPS), DQE (detective quantum efficiency)


Anatomical reference point

(Radiotherapy) The anatomical reference point is an anatomical landmark used for treatment set-up, often using imaging of internal anatomy. It is often chosen to be stable relative to the position of the treatment target, e.g. in treatment of the head and neck, setup may be achieved using x-ray imaging of bony anatomy. The anatomical
Anatomical relationships

(General) Directional anatomical terms describe the relationship of structures relative to other structures or locations in the body.

- **Anterior:** In front of, front (e.g. the kneecap is located on the anterior side of the leg).
- **Posterior:** After, behind, following, toward the rear [e.g. the shoulder blades (scapula) are located on the posterior side of the body].
- **Distal:** Away from, farther from the origin (e.g. the hand is located at the distal end of the forearm).
- **Proximal:** Near, closer to the origin (e.g. the proximal end of the femur joins with the pelvic bone).
- **Superior or Cephalic:** Above, over or towards the head (e.g. the elbow is superior to the hand).
- **Inferior or Caudal:** Below, under or towards the feet (e.g. the foot is inferior to the knee).
- **Medial:** Towards the mid-line, middle, away from the side (e.g. the middle toe is located at the medial side of the foot).
- **Lateral:** Towards the side, away from the mid-line (e.g. the little toe is located at the lateral side of the foot).
- **Contralateral:** On different sides of the midline. The right shoulder and left hip are contralateral to each other.

Anechoic (Ultrasound) Anechoic, or hypoechoic, describes areas of an ultrasound image which contain no visible echoes. This is normally a result of imaging fluid where there is little or no backscattering. An example is shown in Figure A.34.

Aneurysm (General) An aneurysm is an abnormal local enlargement, or dilatation, of a blood vessel. The term usually refers to an artery although locally enlarged veins may also be described as aneurysmal. Aneurysms are found in several arteries though they are most commonly seen in the abdominal aorta. A pseudoaneurysm or false aneurysm is a contained leak from an artery, usually iatrogenic and often as a result of femoral artery puncture during arteriography.

**Clinical Consequences:** Typically aneurysms may grow in size and possibly rupture; the clinical consequence of this depends upon the site. In the brain the bleeding may be into the brain itself or into the surrounding subarachnoid space, which can result in a variety of symptoms including stroke and may be fatal. In the abdominalorta there may be extensive blood loss into the abdomen which is often fatal. Abdominal aortic or cerebral artery aneurysms often have no or non-specific symptoms. It is therefore difficult but important to correctly diagnose their presence. If this is done, an elective procedure has a much higher success rate than emergency treatment after rupture. An aneurysm may become lined with thrombus, throw off emboli or even occlude. An acutely occluding popliteal artery aneurysm can result in critical ischaemia and limb loss.

**Imaging:** Ultrasound is used to screen for abdominal aortic aneurysm; it is a simple, noninvasive test with a high sensitivity. It may be performed as a screening test on those particularly at risk, men aged 65 years and older. Abdominal aortic aneurysms may be imaged with conventional arteriography but if lined with thrombus the true diameter may be underestimated. CT angiography provides detailed images prior to treatment. MRI and MR and CT angiography as well as conventional arteriography may be used to image cerebral artery aneurysms.

**Treatment:** The traditional treatment for an abdominal aortic aneurysm is to replace the affected artery surgically with an artificial graft; this effectively excludes the aneurysm from the circulation. A newer procedure, endovascular aneurysm repair, uses stents introduced via arterial catheters to exclude the aneurysm. A cerebral artery aneurysm may be surgically clipped or embolised with metallic coils introduced via arterial catheter under radiographic control.

**Related Article:** Aneurysm clips

Aneurysm clips (General) Aneurysm clips are metallic, often titanium, surgical devices used to prevent the rupture of intracranial aneurysms. They take the form of a clip with blades and a coiled spring which ensures closure of the blades. The clips are designed to isolate balloon-like aneurysms from the intracranial arterial circulation. The clip is applied under visual control using an operating microscope through a surgical opening in the skull called a craniotomy. The clips work best on aneurysms with a distinct neck that separates the aneurysm from the artery. An alternative to clipping is coil embolisation where small platinum coils are introduced into the aneurysm under radiographic control through an arterial catheter. The coils cause the aneurysm to thrombose, thus isolating it from the circulation.

**Related Article:** Aneurysm


Anger logic (Nuclear Medicine) The event localisation process in a scintillation camera system is referred to as anger logic. In conventional analog
Anger scintillation camera

The scintillation camera was invented by Hal O. Anger (1920–2005) (shown in the following picture) in the mid-1950s and is also referred to as Anger scintillation camera. In some texts also the term ‘gamma camera’ is used.

In a scintillation camera systems the position is determined by splitting the PM-tube signal into four different output lines. The signal of each output line is denoted as \(X^+\), \(X^-\), \(Y^+\), \(Y^-\). Each output line is associated with a resistor and the value of the resistance differs between the different output lines (see Figure A.35). The output line signals are used to determine the \(X\) and \(Y\) position over the entire detector surface.

The \(X\)-position is given by the ratio between the difference in \(X^+\) and \(X^-\) signal and the total \(X\) signal \((X^+ + X^-)\). The same goes for the \(Y\) position:

\[
\begin{align*}
X &= \frac{(X^+ - X^-)}{(X^+ + X^-)} \\
Y &= \frac{(Y^+ - Y^-)}{(Y^+ + Y^-)}
\end{align*}
\]  

(A.23)

The positions are normalised so that the calculated position does not depend on the energy deposited, i.e. pulse height. \(Y\) and \(X\) can range from \(-1\) to \(+1\) and in a perfect scintillation camera the values would change linearly when moving from the lower left corner \((-1, -1)\) to the top right corner \((+1, +1)\). Non-linearities will give rise to either pincushion or barrel distortion.

In digital cameras the signal from each PM tube is digitised and the position is calculated using software. The digital event localisation is analogous to the resistor read out but it also allows for more complicated algorithms. One commonly used approach to improve the positioning accuracy is to discriminate PM tubes with low signals from the position calculation. The largest signal contribution in the positioning accuracy is to discriminate PM tubes with low signal. Another advantage with this approach is that when only a few PM tubes surrounding the interaction position are used for positioning, the other PM tubes can be used for simultaneous acquisition, thus increasing the count rate performance.


Related Articles: SPECT, Barrel distortion, Pincushion distortion, Photomultiplier (PM) tubes

Related Articles: Scintillation camera, Scintillation crystal


Angiogram

(Magnetic Resonance) Angiogram is an image showing vessels, obtained by, e.g. x-ray or MRI. To achieve this result, contrast media are injected into the vessels (e.g. contrast media with Iodine for x-ray angiography, or paramagnetic nanoparticles in MRI). See also Magnetic resonance angiography (MRA) and digital subtraction angiography (DSA).

Related Articles: Magnetic resonance angiography (MRA), DSA

Angle of beam incidence

(Radiotherapy) See Oblique incidence

Ångström

(Nuclear Medicine) An angstrom (ångström) is a unit of length equal to \(1 \times 10^{-10}\) m (or 0.1 nm) and is abbreviated as Å. It is used in the field of spectroscopy, atomic physics and chemistry where the size of atoms, visible light spectra and length of chemical bonds are sometimes measured in ångström.

Angular anisotropy effect

(Nuclear Medicine) See Anisotropy

Angular sampling intervals in computed tomography

(Diagnostic Radiology) On 3rd generation CT scanners the x-ray tube and detectors rotate around the object being imaged, to acquire attenuation data at different angular positions (Figure A.36). The number of ‘views per rotation’ refers to the number of times the detectors are sampled during one tube rotation which determines

Figure A.35 (a) A schematic representation of the 19 PM tubes in a scintillation camera. (b) The output lines from a single PM tube. The PM-tube signal is divided to four output lines by four resistors.

For a detailed description – see Scintillation camera.
Anisotropy

(Nuclear Medicine) Anisotropy is the property of being directionally dependent as opposed to isotropy which is homogeneous in all directions. Examples of anisotropy are the fission fragments in a nuclear reactor. The fission products of $^{239}$Pu are most likely to yield a fission product couple where the lighter nuclei have a mass of 90–100 u and the heavier of 130–140 u.

**Anisotropy**

Computed tomography

**EXAMPLE A:**

<table>
<thead>
<tr>
<th>Rotation Time (s)</th>
<th>Views per Second</th>
<th>Samples per Rotation</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>2000</td>
<td>2000</td>
</tr>
<tr>
<td>0.5</td>
<td>2000</td>
<td>1000</td>
</tr>
</tbody>
</table>

On other systems, the sampling frequency changes with rotation time, so that the number of times the detectors are sampled remains constant and spatial resolution is maintained. See Example B.

**EXAMPLE B:**

<table>
<thead>
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<th>Samples per Rotation</th>
</tr>
</thead>
<tbody>
<tr>
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<td>2000</td>
<td>2000</td>
</tr>
<tr>
<td>0.5</td>
<td>4000</td>
<td>2000</td>
</tr>
</tbody>
</table>

On some CT systems the views per second (sampling frequency) remains constant with rotation time, so that for fast rotation times the number of views per rotation decreases. See Example A. This leads to a reduced circumferential spatial resolution.

**Related Article:** Computed tomography

**Anisotropy**

ائنوسيري (Nuclear Medicine) Anisotropy is the property of being directionally dependent as opposed to isotropy which is homogeneous in all directions. Examples of anisotropy are the fission fragments in a nuclear reactor. The fission products of $^{239}$Pu are most likely to yield a fission product couple where the lighter nuclei have a mass of 90–100 u and the heavier of 130–140 u.

**EXAMPLE A:**

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<td>2000</td>
</tr>
</tbody>
</table>

The life time of the trapped state can be large (up to hundreds of years) if $E$ is large.

Before using a TLD for dose measurement it is necessary to remove electrons and holes from its trapping centres by a heat treatment (Figure A.37b). This procedure is called annealing.

**Abbreviation:** TLD = Thermoluminescent detector.

**Related Articles:** Thermoluminescent dosimeter, Dose, Radiation dosimetry


**Annihilation**

(Nuclear Medicine) The interaction between a beta emitted positron and an electron. As a result of the interaction the particles are annihilated, hence the name of the process. The interaction takes place when the beta particle is brought to a near or complete stop and as a result two photons are emitted from the point of interaction, each photon with an energy equivalent to the electron rest mass of 0.511 MeV. Because of the conservation of momentum the two photons are emitted in an almost exactly opposite direction (Figure A.38).

Beta emitters are used in PET imaging.

**Related Articles:** PET, Beta decay


**FIGURE A.36** Diagram of CT x-ray fan beam at three different angular positions.

**FIGURE A.37** (a) Formation of electron–hole pair and its trapping; (b) removing of electron and holes from traps by annealing.
Annihilation coincidence detection

(Nuclear Medicine) The spatial localisation process in PET is called annihilation coincidence detection (ACD). During the annihilation process a positron undergoes mutual annihilation with an electron which produces two photons with identical energy (511 keV). The photons are emitted simultaneously with an almost 180° opposing direction. The two annihilation photons are registered simultaneously and the system can localise the origin of the event along a line between the two detectors. This line is referred to as the line of response (Figure A.39).

When one photon is detected the coincidence processor examines events in opposite detectors during a specified coincidence timing window, which is typically 6–12 ns. A coincidence is assumed to have occurred when a number of simultaneous events are detected within the coincidence timing window.

When using ACD to localise events, there is no need to use absorptive collimation, which is necessary in SPECT to get any spatial resolution. ACD can be seen like an electronic collimation instead of a more physical approach as in SPECT. Since no spatial resolution is necessary in ACD, the high sensitivity makes PET more suitable for relatively fast dynamic studies and for SPECT or conventional planar imaging. The high sensitivity instead of a more physical approach as in SPECT. Since no spatial resolution is necessary in ACD, the high sensitivity makes PET more suitable for relatively fast dynamic studies and for SPECT or conventional planar imaging.

Related Article: Positron emission tomography (PET)


Figure A.38 Schematic representation of an annihilation interaction between a positron and an electron. Following the annihilation two photons are emitted with a 180° angle to each other.

Annihilation photons in positron decay

(Nuclear Medicine) See Annihilation radiation

Annihilation radiation

(Nuclear Medicine) Annihilation radiation refers to the process when a particle and its antiparticle collide. In this process the two particles are annihilated and radiate photons or particles. In nuclear medicine, annihilation radiation refers to the process where an electron collides with its anti particle, the positron. The positron is usually emitted from a β+-emitting radionuclide like 18F or 68Gd and when brought to a halt it pairs up in an annihilation process with an adjacent electron that emits two photons. In order to conserve momentum and energy the two photons are emitted in opposite directions with total energy equal to the rest mass of the annihilation particles, i.e. 511 keV for each photon.

The fact that the annihilation is followed by emission of two photons with opposite direction constitutes the foundation for PET imaging. After registering an event in a detector the opposite detectors are scanned for a potential detection of the second annihilation photon within a time frame. If such an event occurs, a line of response is recorded between the two detectors.

Related Article: Positron emission tomography (PET)

Annual limit of intake (ALI)

(Nuclear Medicine) Annual limit on intake, ALI, is defined by the ICRP (Publication 30) as the derived limit for the amount of radioactive material taken into the body of an adult worker by inhalation or ingestion. The annual limit of intake (ALI) is the intake of a given radionuclide by inhalation, ingestion or through the skin in a year by the reference man that would result in a committed dose equal to the relevant dose limit. The ALI is expressed in units of activity.

ALI is the smaller value of intake of a given radionuclide in a year by the reference man that would result in a committed effective dose equivalent of 50 mSv or a committed dose equivalent of 500 mSv to any individual organ or tissue.

The ALI of any radionuclide depends on the following factors: (1) the type of radiation emitted, (2) energy of the radiation and that of any radioactive progeny, (3) the selective biodistribution and accumulation in specific organs or tissues and (4) the effective half-life.

Note: ICRP intended to replace ALI with Dose Coefficients with Publication 68 (1995), but ALI is still in use.

Abbreviation: ICRP = International Commission on Radiation Protection


Annular array

(Ultrasound) Annular arrays are transducers with a concentric ring of transducer elements. By altering the time or phase between elements, the depth of best focus can be altered in the imaging plane and in the elevation plane (Figure A.40). The transducer is rotated about an axis to sweep through a volume to produce a sector image. Conventional annular arrays are no longer available on most commercial scanners.

Experimental annular arrays have been used for intravascular imaging. By arranging elements in a ring/annular configuration in a catheter, imaging has been shown to be possible both radially and in a forward direction. A possible configuration is shown in Figure A.42.
Anode

(General) The positive electrode (e.g. of a detector).

Anode acceleration

(Diagnostic Radiology) Anode acceleration is very important for minimising the time between activating the exposure (pressing the button) and the actual radiograph. This time is usually 1–2 s and depends on the motor of the rotation anode, the bearings and the mass (kg) of the anode disc. Almost all x-ray tubes with a rotation anode use an induction electrical motor with several speeds of rotation. The available speeds depend on the frequency of the electricity supplying the motor. All x-ray generators have a special accelerating circuit (anode starting device) initially supplying the motor with higher frequency (to quickly reach the necessary revolutions per minute [rpm]). In high frequency generators this is achieved very easily, as all circuits there use varying high frequency. This acceleration problem does not exist in x-ray tubes with liquid metal bearings.

Related Articles: Anode, Rotation anode, Anode rotation speed, Anode starting device, Bearing, High voltage generator, Medium frequency generator

Anode angle

(Diagnostic Radiology) The anode of the x-ray tube is normally angled; the range of the angle (bevel) is most often between 6° and 20° (Figure A.41). The angle of the x-ray tube determines both the size of the effective focal spot $F_e$ and the size of the actual focal spot $F_t$. The link between these focal spots and the sine of the anode angle $\alpha$ is known as line-focus principle:

$$F_e = \sin \alpha \cdot F_t$$

A smaller anode angle produces a smaller effective focal spot and therefore better spatial resolution in radiographs. However, if the anode angle is very small, then the useful field (over the film/detector), covered by the x-ray beam, will be too small (as the conical x-ray beam will be too narrow). Often x-ray tubes used in CT have a very small anode angle, as they only need to cover a narrow slice and not a large field. Some tubes are made with surfaces forming two Anode angles (see Biangular anode disc).

Some contemporary x-ray tubes (especially mammographic tubes) use anode target perpendicular to the beam of thermal electrons (i.e. no bevel of anode surface). In this case the tubes are mounted in a tilt position (in comparison to the image plane), this way presenting an effective anode angle (incorporating the tilt angle). This requires the anode angle to be defined in relation to the central x-ray beam – the anode angle is the angle between the central x-ray beam and the anode target. This definition satisfies the classical geometry in Figure A.41 (angle $\alpha$), as well as the situation when the x-ray tube is tilted.

Figure A.42 represents three cases: on the left is a normal x-ray tube with bevelled anode (actual anode angle 15°); in the middle is the same tube, but tilted at angle $\beta = 6°$, thus presenting an effective anode angle of 21° (sum of 15° + 6°); the tube on the right is with perpendicular anode (i.e. no anode bevel), but tilted to 25° ($\beta = 25°$), thus acting as x-ray tube with anode bevel of 25°.

Related Articles: Stationary anode, Rotating anode, Target, Line focus principle, Biangular anode disk, Focal spot actual, Focal spot effective, Focal spot

Anode cooling chart

(Diagnostic Radiology) See Anode cooling curve

Anode cooling curve

(Diagnostic Radiology) This is the chart showing the dissipation of the heat stored in the anode with time (in short this characteristic is also known as cooling curve). Usually this chart is presented together with the chart showing the build-up of heat into the anode (anode heat storage charts, showing the tube load time).

The heat imparted to the anode is absorbed by its structures and by its surrounding; due to this reason the heat storage capacity is a very important parameter which is naturally linked with the
Anode heel effect

The x-ray anode generates radiation in all directions (only a fraction of it is at the direction of the patient). At diagnostic energy, this fraction is mainly at direction 90° from the direction of the incident electron beam (anode current) in the x-ray tube. The intensity of the radiation beam towards the patient has significant spatial variation. Figure A.44 (curve 1) presents an example where the maximal intensity of a new x-ray tube (marked with 100%) is at direction 15° measured from the anode surface (this depends on the type of the x-ray tube). There is a notable loss of x-ray beam intensity (up to 50%) at the anode side of the beam. This is due to lesser production of x-ray photons at this direction (mainly due to absorption of the x-rays in the anode itself at the lower end of the target surface). This decreased intensity of radiation at the anode site of the beam (if one looks it from the place of the patient) is known as ‘heel effect’.

The heel effect is more prominent with old x-ray tubes, where all intensity of the beam decreases up to 50%. This overall decrease of beam intensity is primarily due to loss of x-ray radiation inside the cracks on the target surface. These cracks are result of the thermal stress of the target after thousands of exposures (i.e. cycles of heating and cooling). The second example in Figure A.44 (curve 2) presents an old x-ray tube (used ∼10 years at normal daily workload) where the maximal intensity has dropped at a half of new x-ray tube and has shifted to 20° measured from the anode surface. This has produced a much more noticeable heel effect. Note that the ‘age’ of an x-ray tube (measured with the number of exposures produced) depends greatly on the tube type (target surface alloy, construction, etc.), as well as of the power of exposures.

For most radiographs the heel effect is almost unnoticed, as it is covered (absorbed) by the lead diaphragm attached to the x-ray tube. However the effect is seen with large area films (e.g. 35 × 43 cm²) and specially when the x-ray tube is aged. Figures A.45 and

EXAMPLE FROM FIGURE A.43

The heat storage curves represent the anode build-up power in watts and corresponding HU/s. If fluoroscopy is performed with 100 kVp and 3.6 mA, this will result in 1.4 × 360 W = 500 HU/s – for 4 min this fluoroscopy will import to the anode ∼70,000 HU (by the heat storage curve).

If we additionally perform two exposures with parameters 100 kVp, 500 mA, 0.2 s, the total heat delivered by these will be 2 × 100 × 500 × 0.2 W = 20,000 HU. In this case the total heat to the anode will be 70,000 + 20,000 = 90,000 HU. This is safe (in this case the maximum tube heat capacity by Figure A.43 is about 140,000 HU).

The cooling curve shows that these 90,000 HU will be dissipated in 5 min. This is found by subtracting 2 min (where the cooling curve crosses 90,000 HU) from 7 min (where the cooling curve reaches 0 HU).

Related Articles: Tube rate charts, Stationary anode, Rotating anode, Target, Tube load time

Hyperlinks: Sprawls Foundation: http://www.sprawls.org/resources

A.43 Anode heat storage capacity chart and cooling curve. See the example on how to use these curves. (Graphs courtesy of Sprawls Foundation, www.sprawls.org)

cooling. The anode heat storage charts represent the link between the heat units [HU] and the time (minutes) for their absorption. These charts (input curves – HU/s or J/s [W]; 1 HU = 1.4 J) show the heat stored in the anode as a result of a long exposure or a sequence of multiple exposures. These charts are very useful for fluoroscopy. One has to remember that in rotating anode tubes the thermal path (focal spot track) gradually increases its temperature, but in case of a single-phase generator the momentary actual focus (projected focal spot) increases its temperature by pulses in ‘hot spots’ of the thermal path (depending on the anode rotation speed – rpm). The cooling curve on the same chart shows the time (in minutes) for cooling the x-ray tube. The combined chart is presented in Figure A.43.

Sometimes there is an additional cooling curve for the anode housing (which depends on the type of cooling oil in the housing). In contemporary x-ray equipment the tube load/cooling is controlled automatically depending on the cooling curve.

There are special rating charts and heat storage capacity charts for computed tomography x-ray tubes. These are related not only to the focal spot size (as for all tubes), but also to the scanning time. These charts use the same concept, but are for much greater values, as most contemporary CT tubes withstand more than 1 million HU.

A.44 X-ray tube intensity spatial distribution for new (curve 1) and old (curve 2) x-ray tube. The x-ray intensity is shown in relation with the maximum intensity (100%) at the middle of the central beam of a new tube.
Anode (of an x-ray tube) (Diagnostic Radiology) The anode of the x-ray tube is located opposite the cathode at \( \sim 25 \text{ mm} \) distance. In most tubes the anode is angled, the range of the angle (bevel) is normally \( 10^\circ - 20^\circ \) (Figure A.48).

The anode is at positive potential relative to the cathode (in some x-ray tubes the anode is grounded, but the cathode has high negative potential). This way the anode attracts the thermal electrons produced by the cathode. The small region of the anode, which is bombarded by the thermal electrons and produces x-rays is called the target. Almost 99% of the energy imparted to the target by the electrons is converted to heat and secondary electrons are generated (strictly speaking the energy converted directly to heat is \( \sim 75\% \)). Due to this reason the material of the target is normally tungsten – a material with a very high melting point. Tungsten also has a high atomic number, which is important for the effective conversion the energy of electrons to x-rays. This ‘Bremsstrahlung generation efficiency’ (h) of the anode is discussed in the article Target of the x-ray tube.

A.46 (detail) illustrate the heel effect (seen as drop of film optical density) of an old x-ray tube and field \( 24 \times 30 \text{ cm}^2 \).

FFD Dependence: The heel effect is better seen when the focus-film distance (FFD) is shorter, as in this case the lead diaphragm which specifies the radiographic field is naturally more widely opened (wider beam angle); hence, the edges of the beam are displayed. Compare the un-exposed ends of the films in Figure A.47 (upper with FFD = 100 cm and lower with FFD = 150 cm).

Related Articles: Anode, Heel effect

Hyperlinks: EMERALD (DR module), www.emerald2.eu

Anode rotational speed

(Diagnostic Radiology) The speed of x-ray tube anode rotation is proportional to the power of the tube (see Rotation anode). A higher rotation speed leads to a more even heat distribution over the thermal track and therefore improved anode cooling.

Almost all x-ray tubes with a rotation anode use an asynchronous induction electrical motor with several speeds of rotation. At low power x-ray exposures (e.g. Fluoroscopy) the motor is supplied with ~20Hz electricity and rotates with ~1200 revolutions per minute (rpm). When radiographic exposures are made the anode has to absorb more heat. This requires a higher speed of rotation, ~3000 rpm, which is achieved by supplying the induction electrical motor with ~50Hz electricity.

High power exposures (or sequence of exposures in angiography) significantly increase the temperature of the anode, and quicker rotation is necessary to allow its even distribution over the thermal track. In this case the motor is supplied with 150Hz and thus rotates with ~9000rpm. In reality the rpm is a bit less than the earlier-given figure (e.g. ~8500rpm, not the theoretical 9000rpm). Some special x-ray tubes use up to 17,000 rpm.

The speed of rotation is also related to the type of anode bearings. This is of special importance for the initial rotation of the anode. All x-ray generators have special circuitry which allows the exposure only after certain rpm have been reached. Due to this reason the acceleration of the anode rotation is of importance for minimising the time between pressing the exposure button and the actual x-ray generation (very important for targeted quick exposures of moving organs).

Contemporary x-ray tubes with liquid metal bearings have very low friction, and in fact the anode reaches its rpm after switching on the equipment (e.g. in the morning) and constantly rotates (keeping these rpm) during the whole working day/period.

Related Articles: Anode, Rotation anode, Anode acceleration, Cooling curve, Bearing


Anode starting device

(Diagnostic Radiology) See Starting device

Antagonist

(Nuclear Medicine) An antagonist is a molecule that binds to a receptor without activating it. Instead the receptor is blocked and therefore hindered from performing its normal activity. An irreversible antagonist binds permanently with a covalent bond to a receptor. The receptor is then unable to function. An antagonist that binds to a specific subgroup of receptors is called a selective receptor antagonist.

Anterior

(General) Directional anatomical terms describe the relationship of structures relative to other structures or locations in the body.

Anterior: In front of, front (e.g. the kneecap is located on the anterior side of the leg).

Related Article: Anatomical relationships

Anteroposterior (AP) projection

(General) There is a convention where the radiographic technique projection is identified by the direction of the x-ray beam. In the anteroposterior projection the x-ray tube produces an x-ray beam which passes through the back to the front of the patient to produce an image.

Related Article: Technique projection

Anthropomorphic phantom

(Nuclear Medicine) A phantom designed to mimic the morphology of the human body is referred to as an anthropomorphic phantom. Anthropomorphic phantoms are used for

- Evaluation of system parameters in patient-like situations
- Educational use
- Evaluation of artefacts
- Dose evaluation/calculation of delivered dose

Examples of anthropomorphic phantoms are torso (lung, breast and cardiac), skull and pelvis phantoms.

Antibodies

(Nuclear Medicine) Antibodies are proteins found in blood or other bodily fluids. They are a part of the human immune system and they...
identify and target foreign biomaterial, i.e. bacteria and viruses. The antibodies are produced by one type of white blood cell called a B-cell. Most antibodies have a similar general structure, except for a small tip of the protein which is extremely variable. Each of these tips can bind to a specific structure, or an antigen. Antibodies produced by a specific type of B cell are called monoclonal antibodies. These antibodies can be biologically engineered so that they target a specific biophysical process and they are therefore very useful for nuclear imaging and radionuclide therapy.


Antigend toin targetting (Nuclear Medicine) Antigen targeting in nuclear medicine refers to the use of radiolabelled antibodies to target antigens. Antibodies can be molecular engineered to selectively target tumours which are thus ideal for tumour imaging and therapy. For diagnostic purposes, small fragments of antibodies are used because of their favourable biokinetics, i.e. quick target uptake, faster excretion reducing radiation dose and less chance of immunological response. For therapeutic purposes antibodies are labelled with high-energy-emitting radioisotopes (preferably alpha and beta emitters) which accumulate selectively in tumour cells. Alpha emitters with a short half-life, like $^{211}$Bi and $^{212}$At are unsuitable for labelling whole antibodies because the redistribution time for the antibodies widely exceeds the half-life of the two isotopes, hence giving an unnecessary radiation dose to surrounding organs prior to accumulation in the target organ. $^{212}$Bi and $^{212}$At are therefore labelled to antibody fragments rather than whole antibodies.

Related Articles: Tracer kinetic modelling, Receptor targeting, Neuropeptide targeting, DNA targeting, Glycolysis targeting, Apoptosis targeting


Anticoidence circuit in single channel analyzer (Nuclear Medicine) An anticoidence circuit is an electronic circuit which only produces an output pulse if the input pulse occurs at one predetermined level. The circuit will not produce an output pulse if another input is sent simultaneously.

An example of the use of anticoidence circuitry is in radiation detection. A scintillation detector produces a voltage pulse that is proportional to the energy of the incident radiation. These pulses are sorted using a single-channel pulse height analyser where two discriminators $D_1$ and $D_2$ are connected to an anticoidence circuit. Only pulses above a minimum level will be transmitted by the discriminators through to the anticoidence circuit. If these voltages are $V_1$ and $V_2$, respectively (where $V_1 < V_2$), then the following is true:

- If the input pulse is less than $V_1$ it will not be transmitted by either discriminator.
- If the input pulse is greater than $V_1$ but less than $V_2$ it will be transmitted by $D_1$ but not by $D_2$. Therefore the coincidence circuit will produce an output pulse.
- If the input pulse is greater than $V_2$ then it will be transmitted by both discriminators simultaneously. In this situation, the coincidence circuit will not produce an output pulse.

In this way, the pulse height analyser will only produce an output pulse if the input pulse falls between $V_1$ and $V_2$.


Aorta (General) The aorta is the main artery supplying oxygenated blood from the heart to the systemic circulation. It arises from the left ventricle of the heart as the ascending aorta. In the chest it is described as the aortic arch with branches which supply the arms and head. The descending aorta is known as the thoracic aorta to the level of the diaphragm. Inferior to the diaphragm it is known as the abdominal aorta. In the abdomen it finally divides into the common iliac arteries.

Related Article: Aneurysm

Aperture (Ultrasound) The meaning of the word aperture is opening or gap. Theoretically, a plane wave of infinite extent will have the same appearance at all locations in space. Transducers are not infinitely large and therefore cannot produce an infinite plane wave, but they can be thought of as a screen which blocks the infinite plane wave except at the active surface of the transducer. Physically, the image of an opening in a screen where an infinite plane wave can pass is equivalent to a transducer with an active surface that vibrates. The situation is similar in optics, where never an exact geometric shadow of an object is produced. At the edges fringes are produced, an effect that is aggravated as the object gets smaller compared to the wavelength. Christian Huygens visualised this effect as the result of interference from an infinitesimal number of spherical wave radiators on the surface of the aperture. In fact, also the plane wave can be visualised in this manner, but then of course also the aperture is of infinite extent. The result of blocking an infinite plane wave except within a circular surface can be seen in Figure A.50. What is shown is the intensity in the plane that includes the central axis of the beam. The aperture width is approximately four wavelengths. The sidelobes that arise can be reduced by apodization.

Apex (Magnetic Resonance) In filter theory, apodization (Greek de-footing) is the process of reshaping the input signal in order to effect desired changes in its Fourier domain representation and in particular to reduce ‘ringing’ effects. For a temporal signal or a
Apodization

Spatial distribution, apodization is achieved by multiplication by a windowing function. Mathematically this is equivalent to convolution of the Fourier transform of the signal/distribution by the Fourier transform of the windowing function.

In MRI, apodization of SINC (=\(\sin(\pi t)/\pi t\)) excitation pulses can be used to suppress out of slice sidebands resulting from the cut-off at the beginning and end of a pulse of finite duration and to improve the uniformity of excitation through a slice profile. A Fourier transform of a SINC pulse of infinite duration has a rectangular frequency/amplitude profile. In other words a SINC pulse contains a finite band of frequencies of equal amplitude. This is the ideal profile to uniformly excite a slice chosen by slice selection. However a real SINC pulse is of finite duration and includes sharp discontinuities. In the frequency domain representation these discontinuities appear as a ringing effect, distorting the ideal sharp profile of the rectangular frequency profile. By windowing the finite SINC pulse in the time domain these sharp discontinuities may be smoothed out, and a frequency profile closer to the ideal obtained. Common windowing functions used for apodization of symmetric signals include the Hamming and Hanning weighting functions (Figure A.51).

In MRS (magnetic resonance spectroscopy) a Fourier transform is used to generate spectra from FID (free induction decay) signals. The spectra show resonances corresponding to metabolites present in the tissue voxel under examination. Apodization may be used to reshape the FID prior to application of the Fourier transform. Generally the apodization function is a smoothly varying function that attenuates signals occurring later in the FID. The apodization process is equivalent to convolving the metabolite spectrum with a smoothing curve. Apodization produces a spectrum with an improved SNR but with some broadening of spectral lines.

In magnetic resonance spectroscopy imaging (MRSI or ‘chemical shift imaging’) spectroscopic methods are used to generate a low resolution image of the spatial distribution of a metabolite. The relatively small number of \(k\)-space samples in MRSI leads to truncation artefact or ‘Gibbs artefact’ when \(k\)-space data are transformed to an image using a Fourier transform. As a result, data from one voxel can ‘leak’ to other, adjacent voxels. Apodization of the \(k\)-space data prior to suppress outer \(k\)-space values prior to Fourier transformation helps reduce ringing. The apodization process will also reduce spatial resolution.


**Figure A.50** Theoretically calculated beam profile of a circular transducer. The plane shown includes the central axis of the circular aperture. The aperture is actually located a small distance to the left of the figure for computational reasons.

**Figure A.51** Concept of apodization applied to a SINC pulse: (a) SINC pulse and its Fourier transform, (b) truncated SINC pulse and Fourier transform and (c) apodized truncated SINC pulse and Fourier transform.

**Apodization**  
(Ultrasound) Apodization is the distribution of energy across the aperture of a transducer to control the intensity profile of the ultrasound beam and to reduce side and grating lobes. For a single transducer this may be achieved by tapering the electric field towards the edges of the transducer, or by attenuating the beam in the same way over the surface. In arrays the task is simpler, since each element can be controlled individually and thereby the amplitude by which it is excited. Apodization of the receive beam is also used to reduce the effect of side lobes.

The far field pressure magnitude along a cross section of the beam is essentially the Fourier transform of the aperture function. For a rectangular aperture the beam pattern will thus be a sinc-function, and significant side lobes will appear. A strong reflector in the direction of a side lobe can then be interpreted as a weak reflector in the main lobe.

By employing a tapering function on the aperture, as described earlier, the level of the side lobes will decrease. These aperture functions have different names, such as Hamming, Hanning or Gaussian, and all have different properties. The reason for using various functions is that there is a trade-off between several properties of the far field and the aperture function. The side lobe level is probably the most important factor, but has to be weighed against width of the main lobe, as well as roll-off of the side lobes (how fast their amplitude decrease with distance from the main lobe).

**Apoptosis targeting**  
(Nuclear Medicine) This term refers to the targeting of apoptosis enzyme markers by certain radiopharmaceuticals. Apoptosis is self-induced cell death which normally prevents mutated cells surviving. However, this programmed cell sacrifice is lost during the oncogenesis. One example is Annexin V, a protein that targets specific enzymes that are exposed during apoptosis. \(^{99m}\text{Tc}\) or \(^{18}\text{F}\) can be labelled to Annexin for SPECT or PET imaging, respectively. Annexin plays an important role when measuring the apoptosis in different clinical situations, e.g. it could differentiate recurrent tumours from necrosis or measure tumour response.
Related Articles: Tracer kinetic modelling, Receptor targeting, Antigen targeting, DNA targeting, Glycolysis targeting, Neuroreceptor targeting, Hypoxia targeting


Apparent activity

(Radiotherapy, Brachytherapy) Calibration of source strength is a very important part of a comprehensive brachytherapy quality system. Instruments, ion-chambers and electrometers, used for source strength determinations, should have calibrations that are traceable to national and international standards.

Specification of Source Strength for Photon Emitting Sources: Source strength for a photon emitting source can be given as a quantity describing the radioactivity contained in the source or as a quantity describing the output of the source:

1. Specification of contained activity
   a. Mass of radium; mg Ra
   b. Contained activity; Ci, Bq

2. Specification of output
   a. Equivalent mass of radium; mg Ra eq
   b. Apparent activity
   c. Reference exposure rate
   d. Reference air kerma rate
   e. Air kerma strength

Apparent activity is a quantity that can be used for all types of brachytherapy sources. Sources are encapsulated, and contained activity is difficult to determine. For brachytherapy dosimetry, the output of the encapsulated source is the quantity of interest, not the contained activity. The quantity apparent activity, an output specification, has been used as an alternative and is still used, especially for radiation protection applications.

The apparent activity of an encapsulated photon emitting source is the activity of a hypothetical unfiltered point source of the same nuclide that gives the same air kerma rate or exposure rate at the same distance from the centre of the source.

In modern brachytherapy dosimetry, reference air kerma rate or air kerma strength is the quantity used to absorbed dose. Apparent activity is still used in some treatment planning systems to specify source strength. The user of such a system is cautioned to use the same value of the air kerma rate constant as the value used in the treatment planning system, when calculating apparent activity from the measured source output, the reference air kerma rate/air kerma strength. Apparent activity is also used in radiation protection applications.

See Source strength for a full description of specification of source strength.

Related Articles: Source strength, Mass of radium, Contained activity, Equivalent mass of radium, Reference air kerma rate (RAKR), Air kerma strength

Apparent focal spot

(Diagnostic Radiology) See Focal spot

Apparent focal spot

(Radiotherapy) Apparent focal spot or apparent source point or position is relevant to electron beams since they have passed through a scattering filter. This will change the beam from its well-defined collimated shape to one which diverges. The degree of divergence will be energy dependent and will mean that electron beams with different energies will appear to have originated at different source positions.

Source position is an important factor when calculating the change in output factor for extended SSD treatments.

See also Effective source point, Apparent source position and Virtual source position

Abbreviation: SSD = Source to surface distance.

Related Articles: Apparent source position, Virtual source position, Effective source point
Apparent power

(General) Apparent power (sometimes referred to as ‘VA’) is the value derived from the known AC current into, and AC potential across an electrical load, whilst ignoring any phase difference which may exist between the two parameters (Figure A.53):

Apparent power = Potential in Volts \text{rms} \times \text{Current in Amps \text{rms}}

In reality the apparent power is the maximum power that would be delivered into the load, and could only reach this level if the load was resistive. If the load has inductive or capacitive properties, then a phase difference would exist between current and voltage, resulting in a lower real power.

The relative phases of the AC potential and current must also be taken into account where a load possesses reactive properties (capacitance or inductance). In such cases the true power delivered to the load is given by

True AC power

= AC voltage_{\text{rms}} \times \text{AC current}_{\text{rms}} \times \cos (\text{phase angle})

Abbreviations: AC = Alternating current and RMS = Root-mean-square.

Related Articles: Alternating voltage, Alternating current, Direct current, Direct voltage

Apparent source position

(Nuclear Medicine) Apparent source position refers to a certain incorrect positioning of photon counts in scintillation detectors using anger logic, occurring at high count rates. The effect is very prominent when the activity is concentrated in two or more hot spots. In anger logic positioning mode the total PM tube signal received in a well-defined time interval is used in centroid estimations, i.e. estimating the point of interaction. If two or more events occur during this interval the resulting event will be placed between the events. This is more likely to happen if the photon intensity is high. Modern scintillation cameras can separate simultaneous events if they are spatially separated by isolating the involved PM tubes from each other and performing individual centroid estimations.

Abbreviation: PM = Photomultiplier.

Related Articles: Anger logic, Photomultiplier (PM) tubes

Apparent source position

(Radiotherapy) Apparent source position is relevant to electron beams since they have passed through a scattering filter. This will change the beam from its well-defined collimated nature to one which diverges. The degree of divergence will be energy dependent and will mean that electron beams with different energies will appear to have originated at different source positions.

Source position is an important factor when calculating the change in output factor for extended FSD (focus to skin)* treatments.

Applicator

(Radiotherapy) In order to provide a useable electron treatment beam it is necessary to attach an electron applicator (sometimes called cone) to the head of the linear accelerator. These applicators typically come in a range of set field sizes (e.g. 6 × 6, 10 × 10, 15 × 15, 20 × 20 and 25 × 25 cm²).

The applicator is needed because the penumbra produced without it would be clinically unacceptable. This is due to the fact that while some beam shaping is provided by the secondary collimators in the head of the linear accelerator, there is a significant amount of scatter both within the linear accelerator and in the air between the accelerator and the patient. Therefore the applicator collimates the beam and defines it typically at a distance of 5 cm from the patient. Some applicators are also used to provide additional electron scatter thus improving the flatness of the beam.

If fields other than the sizes produced by the applicator set are required, it is common to create an appropriately shaped alloy that can be inserted into the end of the applicator.

Related Articles: Electron applicator, Collimation.

Applicator (brachytherapy)

(Radiotherapy, Brachytherapy) Applicators are used in brachytherapy to position sources in the correct place relative to the target volume. A wide variety of applicators exist, for intracavitary brachytherapy including all varieties, e.g. intraluminal, endobronchial, intravascular, and surface brachytherapy, and for interstitial brachytherapy, e.g. needles and catheters.

Applicators for high dose rate afterloading units must be closed in order to maintain source integrity; the source and its drive cable must never come in contact with bodily fluids. Further, care must be taken when cleaning and disinfecting these applicators to ascertain that they remain dry inside.

Pictures of applicators are shown in other articles: Temporary implant–high dose rate needles (closed), Permanent implant – low dose rate needles for seed implantations (open), Intracavitary brachytherapy – radium applicator and high dose rate applicators for gynaecological treatments, Interstitial implant – high dose rate and low dose rate needles.

For safety reasons, it is imperative that vendor-specific connectors and applicators are used in remote controlled afterloading techniques. But, that is actually not a limitation to the type of application that can be made. Using flexible applicator tubes, for instance, it is possible to fabricate custom-designed applicators. To give an example, Figures A.54 and A.55 show a mould applicator with three channels, used to treat a patient with a maxillary cancer with high dose rate brachytherapy postoperatively. The three channels must go out through the patient's mouth for connection to the afterloader. The mould is put into place in the maxillary cavity via the mouth (the patient had undergone surgery previously, as well as preoperative radiotherapy); Figure A.55 shows the three channels and also the fixation steel wire that goes around the patient's teeth in the upper left jaw.

Related Articles: Brachytherapy, Temporary implant, Permanent implant, Intracavitary implant, Interstitial implant, Image guided brachytherapy

* Focus to skin (FSD) is the distance from the focus of the beam to the patient's skin surface.
A

**From back-target; channels 1, 2.**

**From front**

**Lead markers**

**FIGURE A.54** Mould applicator with three channels seen ‘from the back’. The afterloader specific applicators are white.

**FIGURE A.55** The same applicator seen from ‘the front’ with the three channels going out though the patient’s mouth.

### Apron, lead

**Radiation Protection** See Lead apron

### Arc therapy

**Radiotherapy** Arc therapy uses a standard radiotherapy linac and involves the delivery of the treatment beam in continuous mode as the treatment gantry is rotated around the patient for either x-rays or electron beam. Not all linacs are equipped for arc therapy. The treatment may be a single arc or several arcs. The treatment planning system treats the arced beam as a set of discrete static beams close together in angle.

An application of arc therapy is stereotactic treatment of a small target with a small field applicator in treatment sites such as the brain. Recent developments include the combination of arc treatment with multileaf collimator (MLC) leaf movement in intensity modulated arc therapy (IMAT).

**Abbreviations:** IMAT = Intensity modulated arc therapy, Linac = Linear accelerator and MLC = Multileaf collimator.

**Related Articles:** X-ray therapy, Linear accelerator, Stereotactic radiosurgery

### Ardran and Crooks cassette

**Diagnostic Radiology** A testing cassette developed by GM Ardran and HE Crooks (1968) that is used to measure the potential applied to an x-ray tube (kVp) and the filtration. It contains filters of varying thicknesses and works on the penetrometer principle first developed and demonstrated by Roentgen.

The Wisconsin cassette uses a similar principle.

**Related Articles:** Wisconsin cassette, kV meter

### Arithmetic mean of counts in attenuation correction in SPECT

**Nuclear Medicine** See Attenuation correction in spect using conjugate counting

### Array coil

**Magnetic Resonance** An array coil consists of a single physical coil comprising several RF receiver elements. The individual elements provide the high SNR of a small coil, while the combination of many elements provides a large field of view (FOV) of the anatomy of interest.

The development of array coils has been built on the concept of the phased array, where the electromagnetic and electronic design of the coil is optimised to avoid coupling between coil elements. In this way, individual elements of the coils retain their desirable small coil, high SNR properties.

The integration of the array elements with the available receiver channels in the MR system varies between system designs and has evolved over time. In the simplest concept, a single element within a given array coil may be selected and switched into a single receiver channel. In such a case the elements are used independently and separately to build up a large field of view. In modern systems elements within the array coil are used simultaneously. Furthermore elements in multiple array coils may be combined to effectively form a single, large FOV combined array coil. Each element may have its own dedicated RF channel, or share a channel through multiplexing or electrical combination of signals. For example, elements forming part of a pair CP (circularly polarised) coils share channels in some designs (e.g. an 8-element, 4-channel head coil).

Array coil designs have moved towards increased density of receiver elements in order to increase the SNR achievable, evolving from 2 through 4, 8, 16, 32 and greater numbers of elements. Equally, the increased number of RF channels available on MRI systems has allowed dedication of receiver channels to individual coil elements, with less sharing of RF channels. This allows great flexibility in the coverage and set up of individual examinations, with elements from many coils contributing to the final image, without penalty in data acquisition speed.

### ARSAC

**Radiation Protection** See Administration of radioactive substances advisory committee

### Artefact

**Diagnostic Radiology** The term ‘artefact’ is derived from the latin, *arte factum*, and, in diagnostic imaging, refers to structures in the image that are not a true representation of the object.

A fuller description of artefacts in CT, and the origin of different types of CT artefact, is given in each of the related articles listed in the following.

**Related Articles:** Beam hardening, Cone beam artefact, Helical artefact, Image artefact, Metal artefact, Motion artefact, Partial volume effect (artefact), Ring artefact

### Artefact

**General** In medical imaging artefacts refer to false signal phenomena caused by a number of reasons. Artefacts can be introduced by a number of mechanisms, e.g. faulty equipment (e.g. broken PM tube in a scintillation camera) or patient movement during acquisition. For a radiologist it is important to recognise artefacts and separate them from pathological changes.
Arterial input function (AIF)

The arterial input of tracer is unlikely to be instantaneous and thus described by the Kronecker delta function. However, in practice, the arterial input function (AIF) is of relevance in pharmacokinetic modelling, e.g., in the tracer concentration time curve observed in a tissue-feeding artery.

Ultrasound Artefacts are imperfections in an ultrasound image caused by variations of the tissue properties, multiple reflections of the ultrasound pulse, and the physical properties of the ultrasound transducer. When forming an ultrasound image, an ideal operating medium is assumed as follows:

- The speed of sound in the medium is constant
- The attenuation in the medium is constant
- The beam axis is straight
- Pulses received at the transducer originate only from the beam axis
- An infinitesimally thin beam

Variations from these assumptions lead to artefacts in the image. These are categorised as speed of sound artefacts, attenuation artefacts, reflection artefacts, and beam shape artefacts. Artefacts are very common in diagnostic ultrasound imaging and it is very important that the operator is aware of them. Artefacts can be annoying but they may also give the operator extra information about the examined tissue. An example of this is the reverberation pattern from plaque in a vessel, Figure A.56.

Example of a Speed of Sound Artefact – Range Error: In human tissue we assume that the speed of sound is constant at 1540 m/s when calculating the distance between transducer and reflecting surface using the range equation \( d = ct/2 \). Since human tissue is not homogeneous, the speed of sound differs slightly between different types of tissues. For example, in fat, it could be as low as 1420 m/s. If the speed of sound in the medium is less than the assumed 1540 m/s then the echo will arrive later at the transducer than expected. This means that the target position will be projected further away from the transducer than the real target position. This specific kind of artefact is called Range error.

Other speed of sound artefacts include Boundary distortion, Size errors and Refraction.

Arterial input function (AIF)

The arterial input function (AIF) is the tracer concentration time curve observed in a tissue-feeding artery. The AIF is of relevance in pharmacokinetic modelling, e.g., in the calculation of perfusion-related parameters. Theoretically, the ideal AIF would be instantaneous, i.e., the dose distribution over time is described by the Kronecker delta function. However, in practice, the arterial input of tracer is unlikely to be instantaneous and thus extended in time due to the duration of the intravenous tracer injection and the transport through the circulatory system, i.e., from the injection site, via the heart and lungs, to the tissue of interest.

One important quantity in first-pass measurement techniques is the tissue residue function \( R(t) \), i.e., the fraction of tracer that remains in the tissue at a time \( t \) after an instantaneous arterial bolus input. Retrieval of the tissue residue function is normally not straightforward due to the fact that the arterial bolus input is extended in time. However, the convolution integral expresses the relation between the measured tissue tracer concentration \( C(t) \), the arterial tracer concentration \( AIF(t) \), the tissue residue function \( R(t) \) and the tissue blood flow \( F \):

\[
C(t) = F \left[ R(t) \ast AIF(t) \right] = F \int_0^t AIF(\tau)R(t-\tau)d\tau
\]

By monitoring the tracer concentrations in tissue as well as in an appropriate tissue-feeding artery, the convolution kernel can be obtained by deconvolution. Hence, the blood flow \( F \) and the tissue residue function can be determined.

In brain perfusion imaging by dynamic susceptibility contrast MRI (DSC-MRI), one single AIF site is often assumed to represent all arterial input locations of the entire brain (Figure A.57). This approach is, however, dubious since any arterial dispersion occurring between the site of the AIF registration and the true site of arterial input will introduce an error in the deconvolution-based calculation of cerebral blood flow (CBF) and mean transit time (MTT). In order to reduce such errors, the application of local or regional AIFs has been proposed.

Related Articles: Perfusion imaging, Dynamic susceptibility contrast MRI, Cerebral blood flow, Cerebral blood volume, Mean transit time

Arterial spin labelling (ASL) (Magnetic Resonance) Arterial spin labelling (ASL) or arterial spin tagging (AST) is an MRI technique for generating quantitative perfusion maps using the arterial water spins as an endogenous tracer, i.e. without any exogenous contrast agents. For this, the hydrogen nuclei of the inflowing intravascular arterial water are labelled by applying one or two radiofrequency pulses. Normally, a 180° RF pulse is applied to achieve an inversion of the arterial spins upstream of the region of interest, i.e. the imaging slice. The inversed spins arrive at the imaging slice and enter the tissue by means of water exchange between the blood capillary system and the tissue. Hence, they contribute to a reduced longitudinal magnetisation of the tissue and subsequently to reduced signal in the MR image. Therefore, this effect is directly related to the local microcirculation (i.e. the perfusion or regional cerebral blood flow). ASL perfusion maps can be constructed by subtraction of the tagged/labelled image from a control image (without applying an inversion RF-pulse). The intensity of the difference image is directly proportional to blood perfusion. However, the low contrast-to-noise ratio normally requires averaging of a number of ASL images depending on the imaging sequence used (typically of the order of 50–100 images). ASL sequences generally fall into two main groups of measurement techniques, i.e. continuous or pulsed ASL:

Continuous ASL or ASL Steady-State Techniques: Continuous ASL (cASL) is the original technique of ASL proposed by Detre et al. (1992). The blood is continuously labelled as it passes through a plane proximal to the imaging slice using a train of RF pulses. If an equilibrium is reached, the perfusion parameters are calculated by comparing with the signal from the same, unlabeled slice. The drawback with this technique relates to the amount of RF energy (i.e. relatively high specific absorption rate) delivered, long transit time and considerable magnetisation transfer effects.

Pulsed ASL: The primary difference between pulsed ASL (pASL) and cASL is the labelling technique. In pASL, the arterial blood is labelled in a slab using one short RF pulse, which creates a ‘bolus’ of labelled spins. As an example of a labelling technique, the principle of echo-planar imaging and signal targeting with alternating radio frequency (EPISTAR) (Edelman et al. 1994). This technique consists of two acquisitions (Figure A.58). In the first acquisition (Figure A.58a) all spins are inverted in inversion slab below the imaging slice. After a time delay the inverted blood spins (label) will flow into the imaging slice. In the second acquisition (Figure A.58b) no inversion of the slab below the imaging slice is performed as a control experiment. In order to account for magnetisation transfer effects, the control inversion pulse is applied to a slab distal to the imaging slice. Usually a gap between the inversion slab and the imaging slice is introduced to account for imperfections in the slice selection profiles. Absolute quantification of perfusion with pASL requires a reliable model describing the relationship between the measured difference in longitudinal magnetisation and the perfusion. Other common pASL labelling schemes are flow alternated inversion recovery (FAIR) and proximal inversion with a control for off resonance effects (PICORE). Dynamic or time-resolved ASL, where the bolus of labelled water is followed over time, has also been proposed (Petersen et al., 2006). The weakness of pASL is its lower SNR as compared with cASL and its sensitivity to motion between the two acquisitions. The advantages are lower SAR and less magnetisation transfer effects.

For a recent ASL review, see Petersen et al. (1994).


Artificial neural networks

(General) An artificial neural network is a mathematical or computational model conceptually based on the physical and chemical interconnectivity of neurons in a living biological organism.

The biological neuron is a single cell which is organised with potentially many inputs and one single output and this is replicated by the artificial neuron as a mathematical function capable of receiving many inputs and summing them, often with some weighting factor applied and filtered by a non-linear activation function, to produce an output. The artificial neurons are then organised in layers to form a neural network where the output of one layer acts as an input into neurons in the next layer or even feeding back into the input of previous layers.

A collection of such relatively simple mathematical processing elements has the ability to express complex behaviour through the connections between the elements and the connection parameters. An artificial neural network is an adaptive system that can learn by changing its connection structure based on external or internal information that flows through the network during the learning phase. It therefore can be used in non-linear statistical data modelling or to find and recognise patterns in data or images.

Modern artificial neural networks can be used in radiology to assess images, extract specific features and automatically differentiate between various pathological findings.
As low as reasonably achievable (ALARA) principle
(Radiation Protection) A basic principle of radiation protection set down by the International Commission for Radiological Protection in their 1977 Recommendations (Report 26). This is the principle that exposure to radiation should be kept as low as possible to provide an effective diagnostic image while limiting risk to the patient.

Since changed to ALARP (as low as reasonably practicable) to reflect the cost factors (economic and social) that may be considered by the employer.

For more information, see Optimisation.

Related Article: Optimisation

As low as reasonably practicable (ALARP)
(Radiation Protection) A basic principle of radiation protection set down by the International Commission for Radiological Protection in their 1990 Recommendations (Report 26). It superseded the ALARA Principle (as low as reasonably achievable) to reflect the cost factors (economic and social) that may be considered by the employer.

For more information, see Optimisation.

Related Articles: Optimisation, As low as reasonably achievable (ALARA) principle

ASA
(Diagnostic Radiology) The ASA (American Standards Association) is an old system for photographic film speed (photographic exposure), which became the basis for the current ISO film speed system (used worldwide).

ASA has been introduced by the American National Standards Institute (ANSI) – a private non-profit organisation overseeing various standards for products, services, systems, etc. in the United States. ANSI also coordinates US standards with international standards.

The most widely used system for film speed currently is the one produced by the International Organisation for Standardisation (known as ISO).

Another old system for film speed DIN has been introduced by the Deutsches Institut für Normung e.V. (known as DIN), which in English stands for German Institute for Standardisation.

In principle the photographic-related standards of these systems measure the photographic film sensitivity to light. This way the film speed can be determined from the characteristic curve of the film. While this measure is relatively simple for black/white films (as x-ray films), the colour films require separate curves for blue, green, and red.

The current standard ISO 5800 (from 1987) defines both an arithmetic scale and a logarithmic scale for measuring colour-negative film speed.

The ISO arithmetic scale corresponds to the old ASA scale.

The ISO logarithmic scale corresponds to the old DIN scale.

For example

ASA 100 = DIN 21°
ASA 400 = DIN 27°
ASA 1600 = DIN 33°

Due to this reason the film speed is listed as ISO 100/21°; ISO 400/27°, etc.

Related Article: Characteristic curve

Hyperlinks: http://en.wikipedia.org/wiki/Film_speed

Asymmetric energy window
(Nuclear Medicine) In an asymmetric energy window the photopeak is located off centre in the pulse height analyser window. An example of an asymmetric energy window is one where the upper threshold of the window is located at the centre of the photopeak, a so-called asymmetric low window (or off peak low). The opposite situation, i.e. lower threshold at the centre of the photopeak produces an asymmetric high window (or off peak high). Asymmetric energy windows are used to test a NaI (TI) crystal for possible hydration, to tune the PM tubes or to reveal electronic problems (Figure A.59).

Hyperlinks: www.IAEA.org


Asymmetric fields
(Radiotherapy) Conventional beam collimation uses four field-defining jaws or collimators. One of the two sets of opposing jaws move concurrently to define the field width and the other set defines the field length, resulting in a square or rectangular field which has the field centre coincident with the collimator axis. Independent movements of the jaws allow the generation of rectangular beams which are offset from the central axis (asymmetric beams). It is important to distinguish between the central axis as defined earlier and the beam axis which represents the projection of the centre of a defined collimator opening. For symmetric beams they are coincident but they differ for asymmetric beams (Figure A.60).

The asymmetric collimation produces a nontrivial alteration in average beam energy, absolute radiation output, depth dose and beam profiles. The dose due to the radiation scatter can be separated into collimator and phantom components. The collimator scatter for an asymmetric field is almost the same as for a

Asymmetric fields

Related Articles:

Hyperlinks:

Counts
Energy, keV

Counts
Energy, keV

FIGURE A.59 Schematic representation of a low and high asymmetric energy window.
symmetric field of identical dimension while the phantom scatter component of symmetric and asymmetric fields are different. The asymmetric beam collimation changes the off-axis beam quality as a consequence of using a flattening filter to reduce the dose rate at the photon beam centre. In fact the unfiltered photon beam gives a sharply peaked dose distribution because of the angular distribution of the Bremsstrahlung radiation and the presence of the filter to modify the beam intensity results in a greater beam hardening close to the central axis than in off axis regions. This effect causes a hot spot near the edge of an asymmetric beam far from the central axis. The hot spot reduces in magnitude with depth as the relatively softer beam near the edge of large field is attenuated faster than the harder beam near the central axis. This effect, if not taken into account, introduces a dose error of about 5% for large fields. Asymmetric beams require suitable calculation techniques that consider the variation of dose with off-axis distance. Various dose computation methods have been proposed based on the use of appropriate correction factors.


Asymmetric jaws

(Radiotherapy) Conventional linear accelerators have four jaws to collimate the photon beam and usually two opposite jaws move concurrently to define the width and the length of the irradiation field. The resulting square or rectangular field has the centre coinciding with the beam axis. Modern linear accelerators have collimating jaws that can be moved independently of the corresponding opposed jaws and this permits to block a portion of the field from one side without affecting the opposite jaw setting. Some linacs have one independent jaw, others have two or four independent pairs. The independent jaw option is interlocked to avoid errors in the setting of symmetric fields, in which case the opposite jaws open or close symmetrically. When one of the jaws is closed the resultant asymmetric field has a smaller dimension than the original symmetric field and its centre does not coincide with the beam axis of the symmetric field (Figure A.61). Asymmetric jaws are sometimes used to block off part of the field without changing the position of the isocentre.

The uses of asymmetric jaws in clinic are to

- Eliminate the beam divergence at the junction of adjacent fields by blocking one half of the field along the central axis beam thus improving the dose distribution over the junction
- Simplify the matching of two opposite tangential fields
- Irradiate in an arc therapy a target volume that surrounds a critical organ
- Keep the same treatment centre as the original field
- Keep the same treatment centre as the original field in the set up of a boost field

Although these functions have traditionally been performed by beam splitters or secondary blocking on a shadow tray, the use of asymmetric jaws reduces the needed time for the set up and avoids handling the massive shielding block.

Normally the calculation of dose and isodose distribution in a patient requires basic data which are measured with symmetrical collimators. An asymmetric jaw setting produces changes in the depth dose that must be taken into account by the algorithms used to calculate dose and dose distribution.

Asymmetric screen film

(Diagnostic Radiology) Radiography cassettes generally have identical intensifying screens on each side of a double-emulsion film. However, in some designs the two screens have different characteristics. These might be used with a film that has different emulsion designs on each side to produce two different film-screen combinations within one cassette. An example is where one combination produces a desired contrast characteristic and the other enhances detail within certain exposure ranges.

Atom

(General) An atom is a fundamental unit of matter composed of a positively charged nucleus with negatively charged electrons orbiting around it. The nucleus is composed of both positively charged protons and electrically neutral neutrons. Atoms have
equal numbers of protons and electrons; therefore, they are electrically neutral.

**Elements:** Atoms are classified into elements by the number of protons they contain. Elements are represented symbolically by \( Z \), where \( Z \) is the atomic number (number of protons), \( X \) is the chemical symbol, and \( A \) is the atomic mass number which represents the number of particles (nucleons) in the nucleus. For example, the element carbon has six protons and six neutrons and therefore has the symbol \( ^{12}_{6}C \).

**Isotopes:** An element may be composed of atoms that all have the same number of protons, i.e. have the same atomic number \( Z \), but have a different number of neutrons, i.e. have different atomic mass numbers \( A \). Such atoms of identical \( Z \) but differing \( A \) are called isotopes of a given element.

The term isotope is often misused to designate nuclear species. For example, cobalt-60, caesium-137 and radium-226 are not isotopes, since they do not belong to the same element. Rather than isotopes, they should be referred to as nuclides. On the other hand, it is correct to state that deuterium (with nucleus called deuteron) and tritium (with nucleus called triton) are heavy isotopes of hydrogen or that cobalt-59 and cobalt-60 are isotopes of cobalt. Thus, the term radioisotope should be used to designate radioactive species; however, the term radioisotope is often used for this purpose.

The term nuclide refers to all atomic forms of all elements. The term isotope is narrower and only refers to various atomic forms of a single chemical element.

In addition to being classified into isotopic groups (common atomic number \( Z \)), nuclides are also classified into groups with common atomic mass number \( A \) (isobars) and common number of neutrons (isotones). For example, cobalt-60 and nickel-60 are isobars with 60 nucleons each \((A = 60)\); hydrogen-3 (tritium) and helium-4 are isotones with two neutrons each \((A = Z = 2)\).

If a nucleus exists in an excited state for some time, it is said to be in an isomeric (metastable) state. Isomers thus are nuclear species that have common atomic number \( Z \) and atomic mass number \( A \). For example, technetium-99 \( m \) is an isomeric state of technetium-99 and cobalt-60 \( m \) is an isomeric state of cobalt-60.

**Related Articles:** Atomic mass, Atomic mass unit, Atomic number, Atomic weight, Electron, Elementary particles, Isotope, Isotones, Neutrons, Nucleus, Proton, Radioactivity

**Atomic attenuation coefficient**

(Radiation Protection) The average factor by which individual atoms within a medium attenuate the incident radiation is called the atomic attenuation coefficient.

See further details in the articles on Attenuation and Attenuation coefficient.

**Related Articles:** Attenuation, Linear attenuation coefficient, Mass attenuation coefficient

**Atomic emissions**

(General) The term ‘atomic emissions’ covers all electromagnetic radiation and particles emitted from an atom apart from those emitted from the nucleus when undergoing a nuclear transformation. The only particles emitted are electrons and the electromagnetic radiation is in the optical (ultraviolet, visible and infrared) and x-ray region of the electromagnetic spectrum.

The optical and x-ray emissions arise from electrons decaying from an excited state to a lower energy state with the energy of the radiation corresponding to the difference in the energy levels of the two levels.

Atoms in an excited state can be generated by a variety of means – such as heating, electrical discharge or stimulated by external radiation, e.g. a laser beam.

The emission of an electron occurs when the energy imparted to it (by any of the aforementioned mechanisms) exceeds the binding energy of the electron. When an atom emits an electron, it becomes ionised.

A rather special situation is when the emissions arise from internal conversion or electron capture during a nuclear transformation.

**Related Article:** Nuclear transformation

**Atomic Energy Commission (AEC)**

(General) Many countries have or have had an Atomic Energy Commission (AEC), in order to foster and develop peaceful development of atomic science and technology, promote world peace and improve public welfare. In some countries the commission is still active while in Australia and United States it has been closed or transformed. Australia (1958–1981), France (1945–present), Japan (1955–present), India (1948–present), Pakistan (1964–present), United States (1946–1974).

The United Nation Atomic Energy Commission (UNAEC) was founded in 1946, and the first resolution adopted was calling for the peaceful use of atomic energy and elimination of weapons of mass destruction.

Effort was put on the adoption of a resolution giving the power to the United Nations (UN) to impose controls on atomic development that would not be subject to UN Security Council veto. These controls would allow only the peaceful use of atomic energy. Agreement was not reached on this resolution, mainly because the Soviet Union abstained from the proposal. Debate on the plan continued until 1948, although it was clear that agreement was unlikely to be reached. The commission adjourned indefinitely the debate. In this way ended in practice the activity of the commission in 1948.

In the same year 1946, the US Atomic Energy Commission (US-AEC) was founded. At the beginning US-AEC was given extraordinary power and independency to carry out its mission, including freedom in hiring scientists and professionals. The National Laboratory system was established from the facilities created under the Manhattan Project. The Argonne National Laboratory was one of the first laboratories authorised under this legislation as a contractor. At the beginning, before the establishment of the Nuclear Regulatory Commission (NRC), nuclear regulation was also the responsibility of US-AEC. The US-AEC’s regulatory programme sought to ensure public health and safety from the hazard of nuclear power, without imposing excessive requirements that would inhibit the growth of industry. This was a difficult goal to achieve and an increasing number of critics pointed out the insufficiency of the programme in several important areas, including radiation protection standards, nuclear reactor safety and environmental protection. In 1974 the programme was under such pressure that the Congress decided to abolish the Commission.

**Related Article:** International atomic energy agency (IAEA)

**Hyperlinks:** AEC: www.aec.gov; NRC: www.nrc.gov

**Atomic excitation**

(Nuclear Medicine) Atomic excitation is the elevation of orbital electrons from a lower energy state to a higher energy state. It is described by the Bohr atomic model in which electrons are distributed in shells with different orbital distance from the atomic nucleus. In an unexcited atom the electrons are situated in the lower energy states, i.e. the shells closest to the nucleus. Such a system is referred to as the ground state. An electron can be elevated to...
a higher energy state by absorption of energy, e.g. photons with appropriate energy; hence the atom is excited. The atom may return to the ground state by emitting a photon or an Auger electron with a characteristic energy.

**Atomic mass**

(*Nuclear Medicine*) The atomic mass ($m_a$) is defined as the sum of the proton, neutron and electron rest masses in a specific atom. The atomic mass should not be confused with relative atomic mass, average atomic mass and atomic weight. The atomic or molecular mass is often expressed in unified atomic mass units (u). 1 u is equal to approximately 1.66 × 10⁻²⁷ kg (Table A.4).

**Atomic mass unit**

(*Nuclear Medicine*) The unified atomic mass unit (abbreviated as u) is a unit used to express atomic and molecular masses. 1 u equals 1.66 × 10⁻²⁷ kg.

**Related Articles:** Atomic mass, Atomic number

**Atomic number**

(*Nuclear Medicine*) The atomic number (or the proton number) is the number of protons in the atom nucleus. The atomic number is typically denoted as $Z$. Each element has a specific number of protons, i.e. different atomic numbers. The mass number of an atom is determined by the sum of protons and neutrons in the nucleus. Atoms of the same element but with different mass number are referred to as different isotopes of the same element.

**ATP (adenosine triphosphate)**

(*Magnetic Resonance*) See Adenosine triphosphate (ATP)

**Attenuation**

(*Nuclear Medicine*) When a beam of photons travels through matter, a proportion of these photons will be transmitted. Others will be removed from the beam by photoelectric absorption, Compton scattering and pair production. This process is known as attenuation.

The extent of attenuation depends on the energy of the photon and the density and thickness of the absorbing matter. Consider the setup where a narrow beam of mono-energetic photons of initial intensity $I_0$ passes through an absorber of thickness $x$. The transmitted beam has intensity $I$ which is given by the following equation:

$$ I = I_0 e^{-\mu x} \quad \text{(A.24)} $$

The quantity $\mu$ is the linear attenuation coefficient of the absorber for the particular photon energy and it has the unit of cm⁻¹. $\mu$ is the sum of the photoelectric coefficient, the Compton coefficient and the pair production coefficient.

The mass attenuation coefficient $\mu_m$ is obtained by dividing the linear attenuation coefficient by the density of the absorber $\rho$:

$$ \mu_m = \frac{\mu}{\rho} \quad \text{(A.25)} $$

This has the unit of cm²/g. As the density effect is now factored out, this coefficient depends on the absorber atomic number $Z$ and the photon energy $E$.

An important concept in the design of shielding for radiation protection is the half value layer (HVL). This is defined as the thickness of an absorber required to reduce the intensity of the photon beam to half of its original value.

A nuclear medicine image is affected greatly by attenuation. Areas of radioactivity deeper within the body will appear to have a lower count density than those nearer to the surface. In SPECT the images will appear to have higher activity towards the periphery of the patient. Attenuation correction is difficult to apply because the human body is composed of structures of different densities (e.g. lungs, soft tissue and bone). In modern SPECT systems attenuation correction is achieved through the use of transmission sources or a CT scanner.

**Related Articles:** Linear attenuation coefficient, Mass attenuation coefficient, Pair production, Photoelectric effect, Compton effect

### TABLE A.4

<table>
<thead>
<tr>
<th>Mass</th>
<th>Unified Atomic Mass</th>
<th>Kilogram</th>
<th>MeV/c²</th>
</tr>
</thead>
<tbody>
<tr>
<td>Protons</td>
<td>1.007276</td>
<td>1.6726 × 10⁻²⁷</td>
<td>938,272</td>
</tr>
<tr>
<td>Neutrons</td>
<td>1.008 664</td>
<td>1.6749 × 10⁻²⁷</td>
<td>939,573</td>
</tr>
<tr>
<td>Electron</td>
<td>5.485 799 × 10⁻⁴</td>
<td>9.1096 × 10⁻³ⁱ</td>
<td>0.511</td>
</tr>
</tbody>
</table>
Finally, a factor can be defined which is the average attenuation of incident radiation by a single atom within the absorbing material – it is called the atomic attenuation coefficient.

**Related Articles:** Linear attenuation coefficient, Mass attenuation coefficient, Atomic attenuation coefficient

**Attenuation (Ultrasound)** The intensity of a propagating ultrasound wave is decreased by distance. This phenomenon is called attenuation and is due to scattering, reflection, divergence and absorption of the ultrasound beam energy.

The decrease in ultrasound intensity depends on the tissue properties, the frequency and the distance. If the attenuation is expressed in dB, it has been shown that the attenuation is almost proportional to frequency for most kind of tissues, Figures A.62 and A.63. The attenuation coefficient for a specific tissue can be expressed in dB/cm/MHz. Some measured values of the attenuation coefficient for a number of common tissues are shown in the table.

<table>
<thead>
<tr>
<th>Tissue</th>
<th>$\alpha$ (dB/cm/MHz)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Water</td>
<td>0.02</td>
</tr>
<tr>
<td>Blood</td>
<td>0.15</td>
</tr>
<tr>
<td>Liver</td>
<td>0.4</td>
</tr>
<tr>
<td>Muscle</td>
<td>0.57</td>
</tr>
<tr>
<td>Bone</td>
<td>22</td>
</tr>
</tbody>
</table>

**EXAMPLE**

How much will the ultrasound intensity be reduced when traversing 5 cm at 10 MHz through liver tissue?

Attenuation = $(0.40 \text{ dB/cm/MHz}) \times 5 \text{ cm} \times 10 \text{ MHz}$

$= 20 \text{ dB} = 10 \log \frac{I_2}{I_1}$

$\frac{I_2}{I_1} = 10^{20/10} = 100$

The ultrasound intensity is reduced 100 times through 5 cm of liver at the frequency 10 MHz.

**Related Articles:** Absorption, Divergence, Damping, Intensity, Scatter

**Attenuation coefficient (Radiation Protection)** When a beam of ionising radiation is incident on a material, a number of interaction processes can occur that remove energy from the beam (attenuation). The fraction of energy removed can be related to the distance or length travelled through the material, the mass or density of the material, or the atomic density of the material. The terms used to describe this lost fraction are, respectively, the linear attenuation coefficient, mass attenuation coefficient and atomic attenuation coefficient. More generally the amount of energy removed from the beam may be called the attenuation factor.

**Related Article:** Attenuation

**Attenuation correction (Nuclear Medicine)** In planar imaging using a scintillation camera and in SPECT and PET, images are created by measuring photons emitted from a radionuclide distribution inside a patient. The photon penetrates the patient and interacts within the detector, generating a signal that can be measured and used to calculate a position and thereby an image. However, some photons will also interact with the tissue in the patient. The interaction can be either absorption or scattering with or without an energy loss. Thus,
number of photons will be less as compared to the case assuming the radionuclide in air. This reduction of measured photons is caused by attenuation. The attenuation is exponential and depends on the tissue composition, density and the photon energy. In a complex geometry such as the patient the attenuation is heterogeneous and the magnitude depends on the angle for which the detector is positioned. Since it is assumed that the image reflects the activity of the radionuclide in a particular volume within the patient the presence of attenuation will change this activity estimate. This can result in a false defect or be interpreted as a reduction in metabolism. It is therefore generally desired to correct for this effect by applying an attenuation correction. In practice, this is not easy since in many cases the information of attenuating tissues is unknown. Modern scintillation cameras and SPECT systems can have computed tomography capability so that measurements of the anatomy (and related tissue composition and density) can be made accurately.

**Attenuation correction in PET**

*(Nuclear Medicine)* The attenuation correction is the most extensive correction in PET. For an annihilation event at a depth $T$ inside an object with thickness $T$ to be registered as an event, both photons must pass through the object and be detected by the detector. If assumed that they are both emitted in a direction that allows them to reach the detector, the probability that they will reach the detector is the product of their individual probabilities:

$$P_{det} = e^{-\mu x} 	imes e^{-\mu (T-x)} = e^{-\mu T}$$  \hspace{1cm} (A.26)

$\mu$ is the linear attenuation coefficient and it varies with different tissue types (e.g. bone, lung). The probability of detection is only dependent on the thickness of the object and not the position along the line of response (LOR). In PET an attenuation correction factor $A_{i,j}$ for detector pair $i$ and $j$ can be calculated using a blank image and a transmission image. The blank image is acquired using a rod source ($^{68}$Ga) along the axis direction without a subject in the scanner. The rod source is captured using the gamma camera at projections from many angles. To produce the transmission image the same procedure is then repeated with a subject inside the scanner. The correction factor is given by

$$A_{i,j} = \frac{\text{blank}_{i,j}}{\text{trans}_{i,j}}$$  \hspace{1cm} (A.27)

where $\text{blank}_{i,j}$ and $\text{trans}_{i,j}$ are the counts for $i$th and $j$th detector in each projection scan. The transmission scan should be performed prior to the injection of the radiotracer and it is also very important that the patient does not move between the transmission scan and the examination scan. Patient movement can lead to serious artefacts which can manifest as a high or low regional uptake. Another approach is to acquire the transmission image after the examination scan. In such a case the residual activity can interfere with events from the rod source. Since the spatial position of the rod source is known it is also known which detectors are irradiated. An increase in count rate in the irradiated detectors can therefore be measured. The advantage of this method is that it saves time, thereby decreasing the risk of patient movement. The last approach is to acquire both the transmission and the examination image at the same time. This approach is used to save valuable time. A rod transmission source can contribute to the scattered and random coincidences in the acquired images and that is a serious disadvantage. Modern PET scanners are usually combined with a CT scanner (PET/CT). Attenuation correction is therefore applied using the CT data.

**Related Article:** PET


**Attenuation correction in SPECT using conjugate counting**

*(Nuclear Medicine)* In SPECT imaging the number of photons attenuated depends on the source depth (i.e. the thickness of tissue the emitted photon travels through) and the attenuation properties of the tissue. The transmission $T$ of photons is described by

$$T = \frac{N}{N_0} = e^{-\mu x}$$  \hspace{1cm} (A.28)

where

- $\mu$ is the tissue attenuation coefficient
- $x$ is the source depth
- $N_0$ is the number of photons emitted from the source
- $N$ is the number of photons transmitted through the tissue

Without attenuation correction, sources with identical emission rates but at different tissue depths will register different count rates. One approach to deal with attenuation is to use *conjugate counting*. The technique involves the use of two opposite detectors acquiring simultaneously. Conjugate counting over a full 360° is combined to create a data set equivalent to 180° data acquired with a single detector. A source located close to one of the detectors will give a high and narrow response profile and low attenuation in the first detector and larger in the second. The combination of conjugate counting data is performed using two methods: *arithmetic mean* and the geometric mean.

**Arithmetic Mean:** The arithmetic mean is the summation of counts from along a line of response in the respective detector:

$$\bar{T}_i = \frac{(I_1 + I_2)}{2}$$  \hspace{1cm} (A.29)

This simple summation provides a more accurate representation of the activation distribution but there are still some residual position-dependency, i.e. sources in the centre of the object will have a lower and broader response profile than an identical superficial source. A better way to correct for photon attenuation is the geometric mean.

**Geometric Mean:** Using the geometric mean the effects of source depths are practically eliminated. The combining of the conjugation counts into a geometric mean is described by

$$T_0 = \sqrt{I_1 \times I_2}$$  \hspace{1cm} (A.30)

Consider the arrangement seen in Figure A.64.

For photons directed towards detector 1 (upper) or 2 (lower) the attenuation probability is

$$I_1 = I_{01} \times e^{-\mu x}$$  \hspace{1cm} (A.31a)

$$I_2 = I_{02} \times e^{-\mu x}$$  \hspace{1cm} (A.31b)

The geometric mean of the counts registered by each detector is

$$\sqrt{I_1 \times I_2} = \sqrt{I_{01} \times I_{02} e^{-\mu x/2}}$$  \hspace{1cm} (A.32)
As seen in (A.31) the geometric mean depends on the object thickness $D$ and not on source depths $a$ and $b$. The geometric mean is accurate when correcting for attenuation from a single radioactive source. In clinical cases administered activity is often accumulated in a number of volumes (e.g. different organs). For such cases the attenuation correction is favourably performed with the Chang correction method (see separate article).


**Attenuation correction in SPECT using the Chang method**

(Nuclear Medicine) A problem in SPECT imaging is the fact that the probability of a photon reaching a detector depends on the distance travelled in an attenuating object. Consider two sources, a superficial and a deeper lying source. Even if the two sources have identical emission rates, the number of registered events will be higher from the shallow source because of the extensive attenuation of photons from the deep source.

For a single source, attenuation is corrected using conjugate counting (see separate article). The technique uses two opposite scintillation cameras and by combining the two views into one (geometric mean), the effects of attenuation are minimised. This technique is successful when imaging a point source, but in clinical imaging the source is seldom (or never) a point source. To attain quantitative accuracy in clinical imaging a simple method could be used which involves an *attenuation correction factor* (ACF). The attenuation correction is given by multiplying the projection profiles by an ACF using the geometric mean:

$$\text{ACF} = \frac{1}{e^{\mu a} e^{\mu b}} = e^{\mu D/2}$$  \hspace{1cm} (A.33)

where $\mu$ is the linear attenuation coefficient, which is assumed to be constant for the entire object. $D$ is the object diameter or tissue thickness. Another approach is to use the contours of an initial reconstructed image $f(x,y)$ to estimate the path lengths for each pixel for all projections. The ACF is then calculated for each pixel according to

$$\text{ACF}(x,y) = \frac{1}{(1/N) \sum_{i=1}^{N} e^{-\mu d_i}}$$  \hspace{1cm} (A.34)

where $d_i$ is the attenuation path length for a pixel in projection $i$ (total of $N$ projections) and $\mu$ is the tissue attenuation coefficient (again considered constant). A new corrected image $f'(x,y)$ is created by multiplying the reconstructed image $f(x,y)$ by the ACF $(x,y)$ on a pixel-by-pixel basis:

$$f(x,y) = f'(x,y) \times \text{ACF}$$  \hspace{1cm} (A.35)

This technique is known as Chang’s *multiplicative method*.

A more complex implementation for the Chang method is to do a forward projection of the $f(x,y)$ image (see article Filtered back-projection) in order to attain ‘attenuation’ projections. These projections are subtracted from the original measured projection profiles to form a set of *error projections*. The error projections are reconstructed using filtered back projection to form an error image $f_{error}(x,y)$. With this error image the final attenuation corrected image is

$$f(x,y) = f'(x,y) \times \text{ACF} + f_{error} \times \text{ACF}(x,y)$$  \hspace{1cm} (A.36)

The Chang method is relatively successful in regions where the attenuation coefficient is relatively constant, e.g. head and abdomen. But this method has serious limitations when imaging volumes with high density gradients, e.g. lungs and pelvic region. For these regions it is more suitable to use transmission scans and attenuation maps (see separate article) in the Chang method.

**Related Articles:** Attenuation correction in SPECT using conjugate counting, Attenuation correction in SPECT using transmission scans, Filtered back projection, SPECT


**Attenuation correction in SPECT using transmission scans**

(Nuclear Medicine) A problem in SPECT imaging is the fact that the probability of a photon reaching a detector depends on the distance travelled in an attenuating object. Consider two sources, a superficial and a deeper lying source. Even if the two sources have identical emission rates, the number of registered events will be higher from the shallow source because of the extensive attenuation of photons from the deep source.

Numerous approaches to deal with the decrease in count rate for deep lying sources have been suggested; conjugate counting and Chang method are both discussed in separate articles. Both these approaches have limitations, e.g. the Chang method fails to correct for attenuation effects in volumes with high density gradients. Using transmissions scans can lead to a satisfying correction where other methods fail.

The transmission scan is an additional scan, typically performed on the detector system intended for the emission scan. Projections are acquired with an external source, typically a flood or a line source and in some cases with an x-ray tube and a detector mounted next to the scintillation detectors on the camera.

Data acquisition from two scans are acquired; one without an object (blank or reference scan) and one with an object. The relationship between the blank ($I_{blank}$) and transmission counts ($I_{trans}$) depends on the exponential behaviour of the photon attenuation:

$$I_{trans} = I_{blank} \times e^{-\mu x}$$  \hspace{1cm} (A.37)

where $\mu$ is the linear attenuation coefficient. The natural logarithm of the quotient between the two counts is

$$\ln \left( \frac{I_{blank}}{I_{trans}} \right) = \mu x$$  \hspace{1cm} (A.38)
These projection profiles represent the sum of all attenuation coefficients along a line of response

\[ \mu x = \sum_i \mu_i \Delta x_i \quad (A.39) \]

where \( \Delta x_i \) represents the portion of the line of response that runs through the \( i \)th pixel and \( \mu \) is the linear attenuation coefficient in the \( i \)th pixel. A map with a pixel-specific attenuation coefficient is called an attenuation map. The attenuation map could be used in the Chang method to get a more accurate correction. But the most common use of the attenuation map is in iterative reconstruction methods.

As previously mentioned the transmission scan is acquired using a flood source, line source, multiple line sources, a moving line source or an x-ray tube. An important source characteristic is the energy of the emitted photons. The line source emission energy must differ from the energy used for imaging if the two projections are to be acquired simultaneously, using two energy windows. A radionuclide with long half-life as a line source is preferred or else the source must be frequently replaced. Two suitable radionuclides are \(^{153}\text{Gd} (T_{1/2} = 242 \text{ days}, E_{\text{gamma}} = 97 \text{ and } 103 \text{ keV})\) and \(^{123}\text{Te} (T_{1/2} = 120 \text{ days}, E_{\text{gamma}} = 159 \text{ keV})\).

Consider simultaneous acquisition of the transmission scan, using a \(^{123}\text{Te}\) rod source and emission scan, using \(^{99m}\text{Tc} (E_{\text{gamma}} = 140 \text{ keV})\) with two energy windows over the respective photo peaks. A number of counts in the transmission scan will inevitably be Compton scattered photons emitted by a \(^{99m}\text{Tc}\) radionuclide. This contribution to the transmission counts is referred to as downscatter. Downscatter can be avoided by acquiring the two scans sequentially instead of simultaneously.

**Abbreviation:** SPECT = Single photon emission computed tomography

**Related Articles:** Attenuation correction in SPECT, Attenuation correction in SPECT using conjugate counting, Attenuation correction in SPECT using the Chang method, Iterative reconstruction methods, Downscatter


**Attenuation depth**

*Radiation Protection* The attenuation depth in a given medium is the thickness of matter which is needed to absorb completely all the energy of the incident radiation. The attenuation depth is a finite thickness for charged particles such as electrons (called the range). For electromagnetic ionising radiation attenuation follows an exponential law and attenuation depth is not a useful concept.

**Related Article:** Electron attenuation

**Attenuation equation**

*Radiation Protection* Attenuation of photonic radiation through a material is an exponential process that can be represented by the attenuation equation:

\[ I = I_0 e^{-\mu x} \]

where

- \( I_0 \) is the initial beam intensity
- \( I \) is the final beam intensity
- \( x \) is the distance travelled
- \( \mu \) is the attenuation coefficient of the material (absorber)

**Related Articles:** Attenuation, Attenuation factor, Radiation protection

**Attenuation factor**

*Radiation Protection* The attenuation factor is the factor by which the intensity of a beam of ionising radiation is reduced in travelling through a particular medium. Given an incident radiation beam of energy \( E \), for an attenuating element, \( m \), of thickness \( t \), its attenuation factor, \( f_{att}(m, t, E) \) can be defined by the following relation:

\[ I_{out} = f_{att}(m, t, E) \cdot I_{in}. \]

where \( I_{in} \) and \( I_{out} \) are, respectively, the incident and the exiting radiation intensities.

**Related Articles:** Attenuation, Attenuation coefficient, Attenuation equation

**Attenuation steps**

*Diagnostic Radiology* The test of radiographic contrast ability of one x-ray system requires tools with specific Attenuation steps. A typical tool is an aluminium (or copper) step-wedge (Figure A.65). The absorption and thickness of each step is calculated to produce specific attenuation of the x-ray beam. The radiograph of these attenuation steps produces specific optical densities of the x-ray film – a contrast scale. Such tools are used for quality control of x-ray radiographic and fluoroscopic equipment.

One tool used for testing the contrast scale in fluoroscopy is the Leeds Test Object TO GS2. This tool uses very small attenuation steps to assess precisely the greyscale (contrast scale) of image intensifiers. Figure A.66 shows contrast scale produced with TO GS2 from a digital fluoroscopic system allowing a profile of the contrast steps (density profile) to be presented on the screen. The lack of visible contrast steps in the dark part of the greyscale (on left) shows that this system has intrinsic contrast limitation, which cannot be improved with the ‘contrast’ and ‘brightness’ settings of the monitor.

**Related Articles:** Attenuation, Step wedge, Quality control, Radiography, Fluoroscopy, Image intensifier

**Hyperlinks:** EMERALD: www.emerald2.eu

**Attenuator**

*Radiation Protection* Any material that absorbs or scatters the energy incident in a beam of radiation is called an attenuator.

**Related Article:** Attenuation

**FIGURE A.65** Aluminium step-wedge with 14 attenuation steps made for radiographic quality control. (Courtesy of Ing. A Litchev.)
Auto-fluoroscope

*(Radiation Protection)* This is a computer-assisted contouring technique in which the structure of normal tissue organs and treatment targets can be delineated automatically by means of computer software tools. The principle of the technique is based on differences in CT numbers of different tissue structures when contouring in CT images. When using MR imaging, the technique is based on intensity level of MR signals emitted from different tissues.

**Autocorrelation**

*(Ultrasound)* Autocorrelation is a measure of how well a signal matches a time-shifted version of itself. This is used in signal processing to find periodicity in a signal, e.g. A more strict definition is that the autocorrelation $R(t_1,t_2)$ of a signal $x(t)$, is the expected value of the product $x(t_1)x(t_2)$, i.e. $E[x(t_1)x(t_2)]$. Of special interest is the value along the diagonal $t_1 = t_2 = t$ which is the average power of $x(t)$.

In diagnostic ultrasound autocorrelation is mostly commonly known as the name of an algorithm used to estimate blood flow velocity. The estimate is not the result of a correlation per se, but the calculation of the discrete autocorrelation is used to give an estimate of the phase shift between two successively transmitted pulses, reflected off a moving target.

In the derivation of this estimator, the starting point is an expression that relates the estimated velocity along the ultrasound propagation direction to the phase change between two successive pulses. The end result is an expression that includes the arcus tangent of a quotient between sample values for the imaginary and real parts of the two pulses. It turns out that this can be more compactly written as the autocorrelation evaluated at lag 1, namely as

$$v_z = -\frac{c}{4\pi f_0 T_{prf}} \arctan \left( \frac{\text{Im}(R(1))}{\text{Re}(R(1))} \right)$$

where

- $v_z$ is the estimated velocity along the ultrasound propagation direction
- $c$ is the sound speed
- $f_0$ is the transducer centre frequency
- $T_{prf}$ is the pulse repetition time
- $\text{Im}()$ denotes the imaginary part
- $\text{Re}()$ is the real part
- $R(m)$ is the discrete autocorrelation for lag $m$

Using $N$ lines as an estimate for the autocorrelation function, $R(1)$ becomes

$$\tilde{R}(1) = \frac{1}{N} \sum_{i=0}^{N-2} r(i)r(i+1)$$

where $r(i)$ is the complex (quadrature demodulated) received signal, $x(i) + jy(i)$. The asterisk denotes complex conjugation.

**Auger electron**

*(Radiation Protection)* If an inner (K-shell) electron is removed from an atom through some form of ionisation process, the vacancy will be filled by an electron from one of the outer electron shells. The excess energy will be released from the atom in the form of a photon whose energy is equivalent to the difference in the binding energies between the K- and L- (or other) shell electrons. This is the origin of characteristic x-rays from an x-ray tube.

In a smaller number of ionising events the K-shell vacancy will be filled by an outer shell electron but instead of the excess energy being released as a photon, an outer shell electron is emitted. This is known as the Auger effect and the emitted electron is an Auger electron. The Auger effect was named after Pierre Victor Auger who explained the effect in 1925.

The ratio between the number of x-rays per de-excitation and Auger electrons per de-excitation is called fluorescent yield. The fluorescent yield depends on the atomic number of the disintegrating nucleus and the yield is low for elements with high atomic number and high for elements with low atomic number.

Auger electrons are low energy electrons. Auger spectroscopy is a useful analytical tool to determine the composition of materials. In medicine a number of radiopharmaceuticals, most notably Technetium-99m, emit Auger electrons. These contribute to the absorbed dose in a way that is not well understood.

**Related Articles:** Auger effect, Photoelectric effect, Characteristic x-rays

**Audit, quality audit**

*(Radiation Protection)* An assessment of the systems and procedures and the adherence to those procedures within a department using radiation. An assessment of systems and procedures would check compliance against legal requirements (e.g. national ionising radiation regulations) as well as accepted codes of practice. Assessing adherence to local protocol is achieved through various quality control metrics, such as auditing patient administered doses, image quality and equipment maintenance.

There must be appropriate mechanisms in place to review the results of quality audits, to feed back those results and apply corrective actions, and to ensure the actions are completed.

**Related Articles:** Quality control, Quality assurance

**Auger effect**

*(Radiation Protection)* The Auger effect describes the process in which Auger electrons are produced.

**Related Article:** Auger electron

**Auger electron**

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**Related Articles:** Auger effect, Photoelectric effect, Characteristic x-rays

**Auto-fluoroscope**

*(Nuclear Medicine)* Auto-fluoroscope is an early (1963) concept of multiple-crystal scintillation camera, introduced by Bender and Blau. This early device has used 294 discrete NaI crystals arranged in a matrix of $21 \times 14$ (each crystal $8 \times 8$ cm and 4 cm thick).
**Automatic brightness control**

*Diagnostic Radiology* The two main factors in determining the amount of x-ray output required to produce a diagnostically acceptable image is the size and therefore attenuation of the patient, and the sensitivity of the detector in terms of quantum efficiency – i.e. how well the detector converts x-ray photons to a signal in the image whilst keeping noise minimised.

The system for automatic brightness control (ABC) is used in fluoroscopy and assures consistent brightness of the image produced by the image intensifier (II). The system is also called brightness control.

The ABC produces x-ray images with constant brightness, irrespective of the material which is being x-rayed (i.e. patient thickness or absorption). It consists of a detector (sensing the II output luminance of the II) and a feedback system, which changes the fluoroscopic parameters (kV, mA or both – depending on the type of the system) so that optimal brightness of the fluoroscopic image is maintained. The most often used sensors are photodiodes or other photo sensor, or a photo-multiplier coupled at the output of the II.

In some systems the video signal can also be used to drive the feedback (this should not be confused with the video automatic gain control, which is related only to the imaging system).

The ABC changes the x-ray parameters (kV and mA) according to the transparency of the object and most often the operating panel shows patient with various thickness (Figure A.67).


**Related Articles:** Scintillation camera

**Figure A.68** shows a block diagram of a typical ABC system for x-ray fluoroscopy. In principle it has similarities with the automatic exposure control (AEC) system used in radiography. The ABC includes feedback block C1 (most often including an integrator and comparator), which can take signal either from a dosimeter (ionisation chamber – D2) placed between the anti scatter grid (AS) and the image intensifier, or from a photo sensor (D1) which measures the overall brightness at the output of the II (before the TV camera). Although both systems would have very similar effect, the system using D2 effectively monitors the dose rate, while the system using D1 monitors the image brightness. The ABC will vary the kV and mA of the equipment until the measured signal equals a preset value. The block diagram shows x-ray equipment with a high frequency generator, meaning that the feedback will change the frequency of the DC–AC converter.

The change of the mA and kV can be very simple (i.e. step change of kV plus internal step changes of the mA), but contemporary fluoroscopic equipment use microprocessor-controlled operational characteristics (linear on non-linear dependence of kV and mA). These characteristics change the kV and mA in a way to achieve the required brightness and at the same time to influence other parameters (contrast, dose, etc.) Various manufacturers apply different ABC operational characteristics. For example, SIEMENS names some of its curves as Isowatt (uses high dose), Anti-isowatt (makes compromise between dose and image contrast), minimal radiation (used for paediatric fluoroscopy), high-image-contrast (used in interventional radiology), etc.

The values of the kV and mA are directly linked to the x-ray transparency of the examined object, because the ABC is adjusted to maintain a certain value of the II input dose rate (e.g. 0.3 μGy/s). Decreasing the II light output – i.e. II dose input – immediately leads to increase of the kV, mA or both in order to keep the same II input dose rate. However this is also linked with the II field sizes. When a smaller II field size is chosen (e.g. using the magnification 15′ field of an 30″ II), then the II light output decreases and the ABC boosts the dose up. Each reducing of the visible II field size (magnification, or zoom mode of the II) leads to increase of the dose to the patient in the observed field. This increase depends on the ABC system, but can reach more than five times.

*Related Articles:* Fluoroscopy, Image intensifier, Automatic dose rate control, Automatic exposure control, Brightness control


**Automatic circuit breaker**

*General* See *Circuit breaker*
Automatic collimation control

(Diagnostic Radiology) Some collimators (beam restrictors) used in radiography have a sensor which detects the size of the detector (film cassette) and automatically restricts the beam to this size. This is achieved by exact movement of the lead jaws of the diaphragm (most often the Bucky diaphragm). This system is also known as positive beam limitation device (PBL).

The x-ray fluoroscopic systems use collimator with circular shutter. All these systems have automatic collimation control which changes the diameter of the x-ray filed to match the image intensifier field size.

Related Articles: Beam restrictors, Filter compensating, Diaphragm, Collimator, Image intensifier

Automatic control system

(General) All imaging equipment apply various automatic control systems. One such system controls specific parameter of the imaging chain by varying another linked parameter. For example, – controlling the brightness of the diagnostic image by varying the output dose of an x-ray fluoroscopic equipment (automatic brightness control, ABC). In general an automatic control system includes detector, negative feedback and regulator. In the case of ABC, these are, e.g. photo detector, comparator and kV/mA regulator.

Related Article: Automatic brightness control

Automatic dose rate control

(Diagnostic Radiology) The system for automatic dose rate control is used to control the entrance dose rate in front of an Image Intensifier (II). It comprises of an ionisation chamber (as detector) and feedback system controlling the kV and mA. Most often the system is used to control brightness of the II output image. Typical values of dose rate controls (in front of the II) are between 0.15 and 0.30 μGy/s. For more details see the block diagram in the article on Brightness control.

Related Articles: Brightness control, Image intensifier

Automatic exposure control (AEC)

(Diagnostic Radiology) Most contemporary radiographic equipment is equipped with an automatic exposure control (AEC) system. This system consists of a radiation detector and feedback loop, which interrupts the exposure when certain predetermined x-ray exposure (dose level) is reached. The main purpose of the AEC is to ensure consistent optical density (darkness) of the x-ray film, independent of the overall x-ray absorption of the x-rayed object, by varying the x-ray exposure. In digital systems, which do not produce film, the AEC ensures a relatively constant detector dose and noise, and reduces the likelihood of dose creep. Sometimes this system is called auto-timer.

There are many different types of AEC systems. Often the difference relates to the type of detector used (ionisation chamber, photo-timer or solid-state detector). The AEC with ionisation chamber detectors are widely introduced in the radiographic practice. These often consist of a stand with several detectors, known also as ‘dominants’ or ‘cells’, placed in the patient table or stand, immediately in front of the film cassette. Due to the fact that AEC have been most often used for chest exposures, the widely used configuration is with three dominants, the left and the right approximately at the position of the middle parts of the lungs and the central dominant below them, at the position of the spine. This type of AEC is mounted at the vertical (chest) radiographic stand (Figure A.69).

The AEC is controlled through the operator panel (Figure A.70). It allows choosing different combinations of active detectors; mean optical density (darkening) of the film, etc. Sometimes the AEC operates through the so-called ‘organ-automatic’ or ‘anatomic programming’. This AEC system uses a microprocessor with stored data for the most effective exposure parameters for radiography of each anatomical organ. Quality control of AEC is of great importance, as most of the radiographs today are made using AEC – i.e. there is no other control of the exposure parameters than this automatic system.

The AEC systems sense the dose (dose rate) of the exposure, compare it with the pre-set value, take into consideration the sensitivity of the film/screen combination and on this basis interrupt the exposure after a certain period of time. Due to the fact that the AEC measures all radiation, which reaches it, it also measures scatter radiation from the patient. In order to compensate this, an optical density correction system is used. This system is often connected with a ±D switch at the AEC control panel (often D stands for ‘dunkel’ – ‘darkening’ in German). In the case of mammography such compensation is very difficult. This is due
to the fact that the AEC detector is placed behind the film (to minimise the absorption of the low energy radiation used). In this case the transparency of the object (the ratio between $N_{ex}$ – the number of quanta exiting from the object, and $N_{in}$ – the number of quanta entering the object) depends solely on the absorption of the object. This is so, because of the compression device, which compresses the size of the breast tissue to an averaged thickness to reduce scatter, and improve uniformity of absorption. In this case a special microprocessor system is used to automatically correct the optical density of the film.

Figure A.71 shows two typical AEC systems. The upper AEC feedback (through $C_1$) is the most often used system for chest stands. Its detector $D$ is most often a thin ionisation chamber (or several chambers) placed between the anti scatter grid (AS) and the film-screen combination (S/F). The signal from $D$ passed through $C_1$, which is comprised of an integrator and comparator. This way $C_1$ integrates the summary signal from $D$ (i.e. measures dose per time – dose rate) and compares it with a set value. The set value depends on the sensitivity of the film, the necessary Darkening, etc. When the measured summary value (equal to the overall x-ray exposure) reaches the set value $C_1$ signals the x-ray generator and interrupts the exposure. The block diagram shows x-ray equipment with high frequency generator, but in simple x-ray equipment this can just be an interrupting switch. When this AEC type uses semiconductor detector, it should be placed behind S/F, as otherwise its shadow will be seen on the x-ray film.

The lower AEC feedback (through $C_2$) is the system most often used in mammography. Its detector $D$ can be either semiconductor or a thin ionisation chamber. Usually there are two detectors $D$ with a special absorbent filter $F$ placed between them. This way the AEC also takes into account the energy spectrum of the x-rays (very important for the contrast in mammography). The two signals from $D$ are analysed by a special processor $P$, which supplies $C_2$ with signal corresponding the beam quality (x-ray spectrum). This signal, together with the signal from the foremost detector, is analysed by $C_2$ (which again comprises of an integrator and comparator). This way $C_2$ integrates the summary signal from $D$ and the signal from $P$, and compares these with a set value. The set value depends on the sensitivity of the film, the necessary darkening, etc. When the measured integrated signal (equal to the overall x-ray exposure) reaches the set value $C_2$ signals, the x-ray generator interrupts the exposure.

Related Articles: Automatic dose rate control, Chest radiography, Mammography


Automatic film processor

(Diagnostic Radiology) An automatic film processor is a device that processes radiographic film by transporting it through four steps: development, fixing, washing, and drying as illustrated in Figure A.72. The several factors that affect the processing are automatically controlled. These include the time in the developer, temperature and replenishment of the developer solution.

Automatic frequency control

(Diagnostic Radiology) Automatic frequency control is a system in high frequency x-ray generators, which controls the frequency of the DC to AC converter, as this frequency is directly related to the $kV$ and other parameters of the exposure. If one of these parameters changes during the x-ray exposure, a negative feedback triggers the automatic frequency control, which changes the frequency accordingly in order to keep this parameter constant.

Related Article: High frequency generator

Automatic gain control

(Diagnostic Radiology) Apart from the brightness control and dose rate control systems, which keep the image intensifier (II) light output constant, a second circuit is incorporated in the TV system of fluoroscopic equipment. This system (automatic gain control, AGC) controls the video gain of the Video amplifier. This amplifier is immediately after the TV camera and assures good amplitude of the video signal to the TV monitor of the equipment. The AGC system is of special importance when the image intensifier is connected to the TV camera with fibre optics (hence no light output can be taken from the II output), as in this case it can also play the role of brightness control.

However the amplification of the video signal is also linked with the image noise (hence signal to noise ratio, SNR), as increasing the sensitivity or the gain (amplification coefficient) leads to unwanted increase of the noise amplitude. This way the AGC system is not directly linked to the patient dose, but is very important for the final image quality. Normally the system monitors the signal from the central part of the visible area (dominant) of the II and reacts to

- The maximal (peak) value of the video signal
- The minimal value of the signal
- The mean value of the signal

The AGC has special importance for digital fluoroscopic systems (and DSA), which can be with either peak-sensitive or mean-sensitive AGC. One problem with monitoring the mean (summary) value of the video signal is related to the fact that some specific contrast differences can be omitted. Monitoring the minimal value of the...
video signal can lead to areas of the image with over saturation (too bright). Monitoring the peak (maximal) level of the signal will prevent the monitor from saturation, but some darker parts of the image will become even darker, therefore affecting the overall contrast. Due to this reason most contemporary AGC monitor several sensitive regions of the image.

Related Articles: Fluoroscopy, Image intensifier, Automatic dose rate control, Automatic exposure control


Automatic kV reduction
(Diagnostic Radiology) See Radiographic kV control

Automatic line voltage regulation
(Diagnostic Radiology) Automatic line voltage regulation is necessary to compensate the voltage drop in most x-ray radiographic systems (see the article about Voltage drop). Usually the system regulates the autotransformer output voltage (in classical x-ray generators), or the frequency of the DC–AC converter (of high frequency x-ray generators).

Related Articles: Voltage drop, High voltage generator, High frequency generator

Automatic multiple-sample systems of NaI(Tl) well counters
(Nuclear Medicine) For measurements of multiple radioactive samples as tissues, biopsies and blood from patients or experimental animals a multiple sample system is typically used. The system is usually based on a NaI(Tl) scintillation crystal (however, also semiconductor systems are in use) coupled to a single or multiple channel analyser. One or several isotopes with different photon energies can be analysed. The system is equipped with a mechanical sample changing system in that several samples can be loaded in the machine and then counted in sequence. The results can then be fed into a computer with analysing program.

Related Articles: Sodium iodide crystal, Scintillation detector, Radioactive sample, Well counter

Automatic timer
(Diagnostic Radiology) Automatic timer is a term which is sometimes used instead of automatic exposure control (AEC). Contemporary AEC systems use microprocessors which vary the x-ray exposure in a specific way; however, the early AEC systems control only the duration of the exposure aiming to achieve specific optical density (darkening) of the x-ray film. These systems have used pre-set mA and kV and have only varied the time of the exposure, hence their name ‘automatic timer’.

Related Article: Automatic exposure control

Automatic tube current modulation
(Diagnostic Radiology) Automatic tube current modulation (ATCM) is the method used in CT to automatically adjust tube current for variations in patient attenuation. It is also referred to as ‘mA modulation’, ‘dose modulation’, ‘automatic tube current control’ or ‘automatic exposure control (AEC) in CT’. The purpose of ATCM in CT is to achieve the desired image quality for all patients and at the same time optimise radiation dose.

Different levels of ATCM are available on CT scanners. Firstly, the tube current (mA) may be adjusted for overall patient size and kept constant throughout the scan (Figure A.73a). Secondly, the mA may be adjusted with z-axis position, for each x-ray tube rotation (Figure A.73b). Thirdly, it may be adjusted to account for variations in attenuation throughout a rotation, particularly for differences between the anterior–posterior and lateral (left-right) patient dimensions (Figure A.73c). Most commonly, all three levels of ATCM are combined for greatest effect (Figure A.73d).

Prior to performing a scan with ATCM, information must be available relating to patient attenuation. This is usually obtained from the scan projection radiographs (SPRs), usually referred to by their trade names, such as scanogram, scout view or topogram (Figure A.74). For rotational modulation (Figure A.73c) some systems employ so-called ‘on-line’ modulation where the attenuation in the first 180° of rotation is used to adjust the mA in the subsequent 180°.

![Figure A.73](https://www.impactscan.org)
ATCM not only results in more uniform image quality from patient to patient and along the patient, but it is also a dose optimisation tool as a lower mA is applied in regions of reduced attenuation. However, in order to achieve dose optimisation, the user must select an appropriate image quality level. Different methods are used to specify image quality. Some systems require an input of noise level (standard deviation of CT number), whereas others, a reference mAs relating to a standard patient.

ATCM systems are usually referred to by their trade names such as SmartmA/AutomA, DoseRight, CAREDose or SureExposure.

**Abbreviations:**
ARG = Autoradiography and CCD = Charge coupled device.

**Autoradiogram**
*(Nuclear Medicine)* An image made from placing an object containing a radioactive substance or substances on a photographic plate or film, or by coating the object with photographic emulsion. The image is formed by exposure of the plate, film or emulsion to radiation emitted from the object.

Digital autoradiography is performed by using solid-state detectors or scintillation materials. For scintillators CCDs or image intensifiers are used for detecting the emitted light from the scintillator. High resolution autoradiography can be obtained by electron microscopy autoradiography.

For detection beta particles, conversion electrons or alpha particles are commonly used.

**Abbreviations:**
ARG = Autoradiography and CCD = Charge coupled device.

**Autoradiography**
*(Diagnostic Radiology)* A process which produces image record (on film) of material or tissue which is radioactive. The process can be applied both at macroscopic and microscopic examinations, usually through the use of radioactive isotopes. An example of autoradiography use is to record (in vivo or in vitro) radionuelabelled tissue. Autoradiography (or digital autoradiography) is rarely used.

**Autotimer**
*(Diagnostic Radiology)* See Automatic timer

**Avalanche ionisation in Geiger–Müller counter**
*(Radiation Protection)* The avalanche is formed in gas proportional counters when the electric field is over a threshold value. This threshold value depends on the gas and its pressure, e.g. at atmospheric pressure it is about 10^6 V/m. In this case the electrons and ions created in gas by the ionising radiation are accelerated by the electric field. The kinetic energy of electrons can be greater than the ionisation energy of the gas and the secondary ionisation process will occur. The gas gain (multiplication) process is in the form of a cascade which is called a Townsend avalanche. The relative increase in the number of electrons (dn/n) per unit path (dx) can be estimated using the Townsend equation:

\[
\frac{dn}{n} = \alpha \, dx
\]

where \( \alpha \) is the first Townsend coefficient for the gas (\( \alpha = 0 \) for the electric field below threshold value) (Figure A.75).

In a uniform electric field (parallel plate geometry) \( \alpha \) is constant and the number of electrons (\( n \)) per unit path at \( x = 0 \) increases exponentially with \( x \) corresponding to the avalanche progress:

\[
n(x) = n(0) \, e^{\alpha x}
\]

where
\( \alpha \) is the first Townsend coefficient for the gas
\( n(0) \) is the number of electrons per unit path at \( x = 0 \)
\( n(x) \) is the number of electrons per unit path at \( x \)

Usually a cylindrical geometry of electrodes is used in gas proportional counters. In this case the electric field increases in the same direction as the avalanche progresses. For that reason the avalanche growth is stronger than in uniform electric field.

The charge amplification resulting from the avalanche process results in better signal to noise characteristics in proportional counters than in ionisation chambers.

**Related Articles:** Gas-filled radiation detectors, Proportional counter


**Avalanche ionisation in Geiger–Müller counter**
*(Radiation Protection)* In a Geiger–Müller gas-filled detector, an ‘avalanche’ is caused when the accelerated electrons strike the anode causing the emission of UV (ultraviolet) radiation. These UV
Avalanche photodiode

![Diagram of avalanche photodiode](image)

**Avalanche photodiode** An avalanche photodiode is the solid-state equivalent to a photomultiplier. It is a solid-state detector which is able to multiply the charge of the deposited energy, hence increasing the signal output. Scintillation light enters the diode and interactions with the material create electron–hole pairs. The increase in gain is achieved by accelerating these primary electrons in an electric field and allowing them to excite and create secondary electron–hole pairs (an avalanche of electrons) prior to read out. It is possible to achieve gains of up to several hundreds in read out signal. The gain is strongly dependent on the temperature and applied voltage. The electron multiplication is a function of the interaction position hence introducing a statistical variance in the gain. This variance leads to a lower energy resolution relative to low noise silicon detectors. Avalanche photodiodes are preferably used for detecting low energy photons.

**Related Articles:** Gas-filled radiation detectors, Geiger–Müller (GM) counters, Ionisation chamber, Proportional counter


**Figure A.76** Example of avalanches spreading along the anode in a Geiger–Müller detector.

**Avalanche photodiode**

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**Average absorbed dose**

*(Nuclear Medicine)* The average absorbed dose is the total energy deposited (J) in an organ by ionising radiation divided by the target organ mass, $m_t$ (kg). The average absorbed dose can be used to calculate an effective dose to patients using specific weight factors to determine a future risk of developing cancer. The absorbed dose contribution to an organ stems from self-absorption from radiation originating from the target organ itself and radiation originating in other organs.


**Average dose**

*(Radiotherapy)* The average dose or mean dose is determined by calculating the dose at a large number of discrete points uniformly distributed within the specified volume (planning target volume PTV, critical structure) and calculating the mean of these dose values.

**Related Articles:** Maximum target absorbed dose, Minimum target absorbed dose, Median target absorbed dose, Hot spot

**Average life**

*(Nuclear Medicine)* The average life of a radionuclide (or a particular excited electronic or nuclear level of an atom) is usually discussed in terms of the arithmetic mean, often simply called the mean.

Radioactive decay is governed by probabilities. For any given radioactive atom, there is a constant chance or probability that it will decay in a stated period of time. In a collection of nuclei of the same radionuclide, each nucleus will be completely independent of the other atoms as to when it decays.

Each radionuclide has its own probability – for a radioisotope-131 (131I) nucleus it is one chance in a million of decaying per second (1 in 997,779), whilst the probability for a technetium-99m (99Tcm) nucleus it is one in thirty thousand (1 in 31,155). The mean life, $\tau$, of a nucleus before it decays is therefore directly related to the probability – for an 131I nucleus it is just under a million seconds (997,779 s = 11.55 days) and for 99Tcm nucleus it is 31,155 s (= 8.65 h).

The average life is related to the half-life by $\tau = \frac{1}{\ln 2} T_{1/2}$. In the previous examples, the half-life of 131I is 11.57/1.44 = 8.01 days and for 99Tcm, 8.65/1.44 = 6.01 h.

**Related Article:** Half-life

**Average life time of atoms**

*(Nuclear Medicine)* The average life time of an atom is defined by taking the mean life time of a large population of radioactive nuclides. The radionuclide average life time is not equal to the half-life since the fraction of radionuclides that ‘live’ longer than the half-life can live for several half-lives before decaying. The average lifetime $\tau$ is defined as the reciprocal of the decay constant $\lambda$. $\tau$ is related to the half-life $T_{1/2}$ according to

$$\tau = \frac{1}{\lambda} = \frac{T_{1/2}}{\ln 2} = 1.443 \cdot T_{1/2}$$

**Related Article:** Half-life of radionuclides
Average mass energy absorption coefficient
(Radiation Protection) Mass energy absorption coefficients depend on the energy of the radiation and on the atomic number of the medium. When the interactions occur between a radiation beam which has a spectrum of energies, and/or a medium composed of a number of different elements, calculations may be simplified by the use of an average mass energy absorption coefficient which takes into account an effective mean energy of the beam, and the different atomic numbers of the elements in the medium.

Related Article: Mass energy absorption coefficient

Avogadro’s number
(Nuclear Medicine) Avogadro’s number, $N_A$, or constant is the number of atoms in 12 g of $^{12}$C. Since the number of atoms in 12 g of $^{12}$C is also a mole, Avogadro’s constant correlates the number of atoms (or molecules) in one mole of any substance.

$$N_A = 6.02 \times 10^{23} \text{/mol}$$


Related Article: Mole

Axial (transverse) plane
(General) To describe anatomical planes imagine a person standing in an upright position and dividing this person with imaginary vertical and horizontal planes. Anatomical planes can be used to describe a body part or an entire body. The axial or transverse plane is a horizontal plane that divides the body into the upper (superior) and lower (inferior) regions by running through the midsection of the body.

Related Article: Anatomical body planes

Axial resolution
(Ultrasound) Axial resolution can be defined as the smallest separation of a pair of point targets on the beam axis.

The separation between echoes from adjacent scatterers is shown diagrammatically in Figure A.77.

If the pulse duration is $t_p$ then the interval between the end of the echo from the first scatterer and the beginning of that from the second scatterer is given by

$$\delta t = \frac{2d}{c} - t_p.$$

When $t_p = \frac{2d}{c}$ then $\delta t = 0$ and the echoes are unseparated. The distance $d = \frac{c}{2}$ is half the pulse length.

In practice the axial resolution is dependent on the pulse envelope, the signal/noise ratio (SNR) and signal processing within the scanner. Generally, for conventional B-mode systems, the axial resolution is approximately half the pulse length.

The implications for this can be seen in Figure A.78. A longer pulse, either from using an increased number of cycles, e.g. in colour flow imaging, for a lower frequency, increasing wavelength, results in longer echoes which may not be separated from adjacent tissue interfaces.

Axial resolution may be measured using pairs of wires or filaments at progressively decreasing separation distances or by examining the image from a single wire or filament to measure the axial length from a reflector.

Related Articles: Bandwidth, Backing

FIGURE A.77 Axial resolution depends on the ability of the scanner to separate echoes from scatterers in the beam direction. This in turn depends on the length of the first echo. If the echo from the second scatterer arrives before the end of the first echo, then there is no separation of echoes and the scatterers are not resolved.

Pulse length and axial resolution

Interfaces

1
2

Transmitted pulse
Reflected pulses

Incident pulse I
Reflected pulse $a$-from 1st interface, $b$-from 2nd interface
Reflected pulses are separated with short pulse (upper) not separated with long pulse (lower)

FIGURE A.78 A shorter pulse may resolve echoes from adjacent interfaces that in a longer pulse merge together.

Azimuthal
(Ultrasound) The word is of Arabic origin and means ‘the ways’, that is the ways, or directions, a person faces. In navigation it is the horizontal angle in degrees, from true north to the present course, or direction (known as bearing). In the context of ultrasound fields, the term seems to have been used in somewhat differing ways. Usually the azimuthal plane refers to the image plane, and the azimuth steering angle is the angle the ultrasound beam is steered off the normal to the transducer face (as for instance as in a phased array). This resembles the navigation analogy, as to what ‘bearing’ the ultrasound beam has, relative to the normal direction.

Others refer to the azimuthal direction as the elevation direction, that is the direction perpendicular to the imaging plane. Evidently the word should be used together with a definition of its usage in the present context.
**B**

*B₀* (Magnetic Resonance) The static magnetic field strength in an MRI system is conventionally indicated with *B₀*. It is usually expressed in units of Tesla (1 T = 10⁴ G). The *z*-direction is commonly chosen along the direction of *B₀*, although exceptions may occur in open MRI systems. In current MRI systems, *B₀* has a constant value over time, ranging from 0.02 to 3 T. Currently, experimental MRI systems have a field strength of up to 11 T and in MRS, field strength is usually indicated by the proton frequency, for example 600 MHz for I4 T.

*B₀* gradients (Magnetic Resonance) The potential of NMR as the basis of a medical imaging technique was realised by Kudravcev in 1960 and notably by Damadian in 1971, who patented the idea. However, no efficient means existed of mapping NMR signals spatially, an obvious prerequisite for imaging.

In 1952, Carr had demonstrated use of a magnetic field varying linearly in space in the context of NMR spectroscopy. This concept of a magnetic field gradient was to form the basis of spatial encoding, allowing NMR to be transformed into a powerful imaging technique. The technique was developed in the 1970s by Lauterbur and Mansfield, who in 2003 shared the Nobel Prize in Medicine for their work.

The resonance frequency of nuclear spins in a static magnetic field of amplitude *B₀* is given by the expression \( \omega_0 = \gamma B_0 \). Imagine now that the amplitude of the field is made to vary linearly along, for example the *x*-axis. It follows that the resonance frequency will also vary linearly with *x*, according to the following equation:

\[
\omega(x) = \gamma (B_0 + G_x x)
\]

where \( G_x \) is a measure of the steepness of the field variation in *x*-direction (the ‘gradient strength’), usually expressed in units of mT/m.

A magnetic field is a vector quantity, and it is important to note that the direction of the field remains that of the main static field, *B₀*, usually along the *z*-axis (the crano-caudal direction with respect to the patient). Figure B.1 illustrates magnetic field vectors along the *x*-axis without (a) and with (b) imposition of a gradient. In the latter case, the field remains oriented along the *z*-axis but varies linearly in amplitude along *x*. The term gradient direction indicates the direction of field variation (the *x*-axis in this example).

Magnetic field gradients are applied for short intervals of time, on the order of milliseconds, during an MRI pulse sequence. If a gradient is applied while there is magnetisation in the transverse plane, the precessional frequency of this magnetisation varies linearly with position along the gradient direction. When the gradient is switched off, magnetisation returns to a common precessional frequency determined by *B₀*. However, the period of differential precession results in linear variation in phase along the gradient direction, and this variation persists until removed or modified by another gradient. These variations in frequency and phase are the basis of slice selection, frequency encoding and phase encoding, the three principal methods of spatial localisation in MRI which are described in detail in specific articles.

In addition to their role in spatial encoding, magnetic field gradients are also used to eliminate unwanted magnetisation (‘spoil’) in order to avoid generation of spurious echoes. In diffusion-weighted imaging, intense magnetic field gradient is used to introduce diffusion sensitisation into a pulse sequence.

Physically, gradient fields are generated by passing currents through gradient coils built into the scanner (Figure B.2). These are resistive wire coils, with a separate coil designed to generate a linearly varying field along each of the three Cartesian axis inside the scanner bore. In the early days of MRI, gradient coil designs drawn from classical electromagnetism were used: Golay coils for the *x* and *y* gradients and a Maxwell pair for *z*. Today, gradients with superior performance in the context of modern scanners with wide bores and shorter magnets are designed by computational methods.

Gradients along any arbitrary oblique axis can be generated by passing currents through two or three of the gradient coils simultaneously, with amplitudes chosen to yield the required net gradient direction and amplitude.

The gradient set should ideally generate linear field variation along the appropriate Cartesian axis, without any field

![Figure B.1](image1.png)  
**FIGURE B.1** Effect of a magnetic field gradient of static field amplitude.

![Figure B.2](image2.png)  
**FIGURE B.2** Gradient coil designs for the *z*-axis (a) and for *x* and *y*-axis (b).
components in orthogonal directions. However, electromagnetic theory shows that this is not possible, and there are inevitably orthogonal fields known as Maxwell terms. These sources of image artefacts are minimised by careful design, but cannot be completely eliminated.

Gradient performance is specified in terms of the maximum achievable amplitude (the ‘gradient strength’) and the speed with which the gradient can be switched on and off (expressed as ‘slew rate’ in mT/m/s). These are limited for safety reasons, as electrical currents induced in the body can lead to peripheral nerve stimulation, an unpleasant and, in extreme forms, painful phenomenon. Another important parameter is the linearity of the gradient over the desired imaging volume, which ensures correct geometrical representation of the imaged object. Linearity is more difficult to achieve in more compact magnets and particularly in open scanners where flatted gradient sets are required.

**Related Articles:** Diffusion imaging, Frequency encoding, Phase encoding, Slice selection, Spoiling, Spurious echoes

$B_0$ homogeneity

(Magnetic Resonance) $B_0$ refers to the main magnetic field of the MR system, typically varying from 0.1 to 3.0 T for clinical MR systems. The homogeneity of the static magnetic field is an important measure of the quality of the magnet. Inhomogeneities can be produced by the scanner and the magnetic susceptibility of the object being imaged.

The homogeneity of the main magnetic field is measured in parts per million (ppm) within a given spherical volume in the centre of the magnet, the size of this volume given as the diameter of the spherical volume (DSV). Homogeneity requirements for imaging are generally lower than for MR spectroscopy, but for most imaging techniques, homogeneity must be maintained over a large region for good imaging quality. The field should be homogeneous in the range of a few ppm over a 30–50 cm DSV.

**Abbreviations:** DSV = Diameter of a spherical volume, PPM = Parts per million, SNR = Signal-to-noise ratio and T = Tesla.

**Related Article:** $B_0$ inhomogeneity

$B_0$ inhomogeneity

(Magnetic Resonance) $B_0$ inhomogeneity is the degree of inhomogeneity of the main, static magnetic field ($B_0$). Manufacturers try to make the magnetic field as homogeneous as possible, in particular at the isocentre of the magnet, but it is not possible to have a perfectly homogeneous magnet. Homogeneity of the main magnetic field is measured in parts per million (ppm).

**Abbreviations:** PPM = Parts per million and SNR = Signal-to-noise ratio.

**Related Article:** $B_0$ inhomogeneity

$B_1$ (Magnetic Resonance) In an MR system, the radio frequency field strength is conventionally indicated by the symbol $B_1$. To affect a nutation, it is applied at Larmor frequency, usually in a plane transverse to $B_0$. If the strength of the $B_1$ field increases, also the frequency of the $B_1$ field increases linearly. For example, the frequency of the $B_1$ field for a 4 T system is 171 MHz for proton imaging. At such high frequency, the interaction between the $B_1$ field and the human body can no longer be neglected. This interaction is caused by the dielectric resonance since the effective wavelength of the $B_1$ field is now comparable with the dimension of the human body or some body structure. This interaction degrades the $B_1$ field homogeneity and subsequently the quality of images. This interaction can also produce an increase of temperature in some part of the body as brain or eyes because the electric field associated with the $B_1$ field increases with the $B_1$ inhomogeneity and consequently increases the Specific Absorption Rate.

**Related Article:** Specific absorption rate

$B_1$ homogeneity

(Magnetic Resonance) $B_1$ refers to the field produced by the MR RF system. RF system homogeneity can vary greatly as it mainly depends upon the RF coil used. There are two main factors which affect $B_1$ homogeneity, the first is the interaction between the RF field and the object being imaged and the second is the inherent inhomogeneities of the RF coils used in the system.

Volume coils such as the birdcage and saddle coils, produce relatively good $B_1$ homogeneity at their centre. The saddle coil will produce very good $B_1$ homogeneity in the direction of its long axis and the bird cage coil produces the highest $B_1$ homogeneity over most of the coil volume, giving excellent image uniformity. Surface coils are renowned for their inhomogeneity, producing a high SNR at the surface of the object, which rapidly decreases with depth in the object.

Flip angles in conventional RF pulses are sensitive to $B_1$ inhomogeneities; therefore, to remove the effects of $B_1$ inhomogeneity, adiabatic RF pulses can be used. They are amplitude- and frequency-modulated pulses that are mostly insensitive to the effects of $B_1$ inhomogeneity. The disadvantage of these pulses is that they can require longer scanning times.

**Abbreviations:** RF = Radio frequency and SNR = Signal-to-noise ratio.

**Related Articles:** RF uniformity, $B_1$ homogeneity

$B_1$ inhomogeneity

(Magnetic Resonance) $B_1$ refers to the RF system used in MRI. $B_1$ inhomogeneity is one of the most important causes of image non-uniformity. There are two causes of $B_1$ inhomogeneity: firstly, the interaction between the RF field and the object being imaged and secondly, inherent inhomogeneities of the RF coils used in the system.

More detail is provided in the article $B_1$ homogeneity.

**Abbreviations:** NMR = Nuclear magnetic resonance and RF = Radio frequency.

**Related Articles:** RF uniformity, $B_1$ homogeneity

$B$-lines

(Ultrasound) A $B$-mode (brightness of greyscale) image is produced by the data from a large number of $B$-lines, where each $B$-line is echo amplitude data from one pulse-echo sequence. The width of each line will determine the lateral resolution of the image. The number of lines used to produce a ultrasound image affects the frame rate, Figure B.3.

**Related Articles:** $B$-mode, $B$-scanner, Frame rate

$B$-scan

(Ultrasound) A $B$-mode image is produced by the data from a large number of $B$-lines. The word scanning is used for the method to send and receive pulses sequentially along parallel beam lines starting at one end of the array transducer, Figure B.4.

**Related Article:** $B$-mode

Back pointer

(Radiotherapy) The back pointer of a radiotherapy treatment machine, such as linear accelerator, is a mechanical or optical device used to assist beam setup a radiotherapy treatment. The front
Background

Background (Nuclear Medicine) In nuclear medicine imaging, background refers to the signal generated by radioactive sources other than the source of interest. Background signal is also referred to as (detector) noise. For example, signals originating from detector electrical noise and background radiation are considered to be “background”.

Background signal decreases the spatial resolution and image contrast of the imaging system and introduces a source of error in quantitative measurements. For special noise-sensitive applications such as dynamic studies, the background contribution can be subtracted from the accumulated image.

Related Article: Signal-to-noise ratio (SNR)

Background equivalent radiation time (BERT)

(Radiation Protection) The background radiation equivalent time (BERT) is a unit of measurement of radiation dose. Sometimes it is important to give an idea about the magnitude of the doses related, for example to diagnostic radiological investigations, in comparison to the value of background irradiation to which population is exposed.

BERT values are, of course, only a comparative indication, since there is no evaluation of the specific methods and parameters used for the x-ray investigation, as well as of the local–individual background values.

Typically, BERT values given vary from 2 BERT for a dental x-ray to 400 BERT for a barium enema investigation.


Background radiation

(Radiation Protection) There are two kinds of ionizing radiation sources: natural and man-made. The natural sources produce the background radiation. The natural background consists of the cosmic radiation, the terrestrial radiation and the internal radiation.

Cosmic Radiation: All living things on the earth are constantly bombarded by radiation from the sky. Charged particles from the sun and the stars interact with the atmosphere and the magnetic field and produce a shower of radiation, typically beta and gamma radiation. Elevation and magnetic field clearly influence the cosmic radiation; therefore, there are different values in different parts of the world. It is evaluated that cosmic radiation contributes 8% of the total radiation, including the man-made component.

Terrestrial Radiation: Radioactive substances are in nature, namely, soil, water and vegetation. The major contribution to terrestrial radiation is given by uranium and its decay products: thorium, radium and radon. Some of these materials might be ingested with food while radon is inhaled. Locations with higher concentration of uranium and thorium in the soil will have higher terrestrial radiation; therefore, different parts of the world have different values. It is evaluated that terrestrial radiation contributes 8% in addition to the radon that contributes 55% of the total radiation, including the man-made component.

Internal Radiation: All people have radioactive potassium-40, carbon-14, lead-210 and other isotopes inside their body from birth. Again, the individual quantities can vary but the variation is minor compared to cosmic and terrestrial radiation. It is evaluated that internal radiation contributes 11% of the total radiation including the man-made component.

The background radiation accounts for about 81% of all public exposure, while the man-made part accounts for about 19%. Given the earlier-described differences, the average, individual, total radiation dose is evaluated to be circa 3.6 mSv/year. Natural and artificial radiation doses produce the same kind of effects.


Hyperlinks: NRC: www.nrc.gov; IAEA: www.iaea.org

Background signal

(Nuclear Medicine) See Background

Back ing material

(Ultrasound) PZT is used as transducer material because it can be moulded in different shapes and can efficiently convert electrical energy to mechanical energy and vice versa. However, the mismatch in acoustic impedance between PZT (30 x 10⁶ kg/m²/s) and
soft tissue ($1.6 \times 10^6$ kg/m$^2$/s) is a significant disadvantage for the transmission of ultrasound energy into tissue and for the ability of transducers to produce short acoustic pulses necessary for high-resolution imaging. The mismatch will cause internal reverberations within the transducer element.

A damping backing material, mounted on the backside of the transducer element, with an acoustic impedance adjacent to the acoustic impedance of PZT, will reduce the $Q$-value (ringing) effectively. Unfortunately, this will also reduce the sensitivity (Figure B.5). The most efficient way of reducing the $Q$-value without reducing the sensitivity is to use a matching layer at the front. A common compromise is to use a backing material with acoustic impedance slightly lower than that for PZT.

The backing material should ideally damp out, or absorb, all the ultrasound energy so that none of the ultrasound energy will be reflected back into the transducer element.

In continuous wave applications, as CW-doppler, high $Q$-value transducers are required and therefore air backing is commonly used. Air will guarantee total reflection on the backside of the transducer disc, which will minimise the ultrasound energy loss.

**Abbreviation:** PZT = Lead zirconate titanate.

**Related Articles:** Transducer, Matching layer, PZT

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**Back-projection**

*(Diagnostic Radiology)* See Back-projection reconstruction

**Back-projection reconstruction**

*(Diagnostic Radiology)* Many medical imaging modalities produce projection data: CT is the archetypal example, and PET is also a source of projection data. The first MR image from P. Lauterbur was also a projection (MR radial sampling). The great insight for the revolution of medical imaging in the 1970s was the realization that it is possible to reconstruct images from such projections.

Before describing how these reconstructions can be done, a remark on the test images: The classical test object for testing projection reconstructions is the Shepp–Logan phantom. This is a synthetic image made of ellipses. The projection transform and Fourier transform of ellipses can be computed analytically; thus, the entire transform of the Shepp–Logan phantom can be computed to arbitrary precision.

**Back-Projection:** The plain back-projection does not give good reconstructions but helps understand what happens for the more correct ones. The idea is very simple: for each projection angle, we smear back the values on the image domain. The full back-projection is the average of these images.

Consider the following $4 \times 3$ image $s$, with an object in the middle Figure B.6. We also show the projections:

$$p_\theta = P(s, \theta = 0, 90)$$

**Fourier Transforms of Projection, the Fourier Slice Theorem (FST):** The Radon transform, i.e. the projection operation, is a transform from image space to a space of lines in the sense that we get one value per line. A line is specified by a direction $\theta$ and a distance $r$ from origin. It can be related to the Fourier transform, the ‘Swiss army knife’ of image processing. It is possible to perform a Fourier transform on the distance variable. The Fourier slice theorem, or projection slice theorem, relates the 1D Fourier transform of a projection to slices of the 2D Fourier transform of the image.

The FST claims that in the following figure (Figure B.7), we get the same result by following the upper path (2D Fourier followed by slice extraction) or the lower path (projection, slice extraction, then 1D Fourier).

**FIGURE B.5** The role of backing material to produce short ultrasound pulses. (Graphs courtesy of EMIT project, www.emerald2.eu)

**FIGURE B.6** Example of plain back-projection. $\mathcal{P}$, projection operator; $\mathcal{B}$, back-projection operator.
**Filtered Back-Projection:** The filtered back-projection algorithm addresses the drawback of back-projection: the back-projection blurs the image Figure B.8. This is because the back-projection is an intuitive operator, but is ‘not’ the mathematical inverse of the projection. A more careful investigation of the transform using the Fourier approach shows that the correct inverse involves a filtering, then a back-projection. Indeed, as the FST tells us what the Fourier transform of projections are, we can use inverse Fourier transformation to recover the image (Figure B.7 shows this clearly, to invert the projection, the arrow $P$, we can follow the dotted arrows: $S1 \rightarrow \text{DFT1} \rightarrow IS2 \rightarrow \text{IDFT2}$). However, remember that the variable is a ‘distance’ as in a polar/spherical coordinate system and the formula for integration in polar coordinates involves an integrating factor proportional to that distance, which acts as a filter. Theoretically exact inversion would thus require computing for all values of this distance, but finiteness and discretisation requirements result in different choices of filtering.

Note that the projection is called Radon transform in mathematics, and we thus want to compute the inverse Radon transform. Equivalent formulas for projections from lines in space are even more complex. Recently, Katsevitch developed 3D formulas for helical reconstruction, which generalise the Radon formulas.

**Iterative (See Iterative algorithms):** The Radon transform, and related projection transforms, are at least approximately linear operations. This implies that rather than using specific discretised version of the continuous inverse formula, we can try to use techniques from the inversion of linear operators. The modern toolbox of linear inversion contains a wide range of iterative techniques.

A very simple iterative idea for the solution of linear systems is the following: every row of a linear system of equation specifies a linear space (or hyperspace in higher dimensions). We can pick any starting point, find the nearest point on the first space, then the nearest of this on the second space, and iterate. This algorithm (Kaczmarcz) is well suited for reconstruction of projection. For projection data, the ‘nearest’ involves a back-projection, and the corresponding algorithm belongs to the class of algebraic reconstruction technique (ART).

**Related Article:** Iterative algorithms


**Hyperlink:** [http://www.maths4medicalimaging.co.uk](http://www.maths4medicalimaging.co.uk)

**Backscatter**

*(Radiotherapy)* At kilovoltage energies, the dose at the surface of an irradiated phantom (or patient) is greater than the dose at the same point if no phantom is present. This is particularly true at energies below 400 kV and arises from the fact that phantom materials scatter a fraction of the primary radiation back to the surface. This contribution to the dose at the point in the phantom is known as backscatter.

**Related Article:** Backscatter factor


**Backscatter**

*(Ultrasound)* Ultrasound is reflected off large surfaces and tissue interfaces. Within tissue, ultrasound is scattered from small-scale variations in acoustic impedance. In this context, small means smaller than the wavelength. In this case, the sound is not re-directed in a specific direction but rather in many directions. The energy scattered in various directions is given by the expression for the scattering cross section, which is a function of angle, the angle relative to the propagation direction of the incident. The sound waves reflected at an angle of 180° are known as backscatter. The backscattered sound is what is detected in diagnostic ultrasound and contributes to the ultrasound image.

**Backscatter factor**

*(Radiotherapy)* The backscatter factor is defined as the ratio of an appropriate radiation quantity (e.g. dose or exposure) at the reference point, i.e. on the surface of the patient or phantom on the central axis of the beam, to the equivalent quantity at the same position in the absence of the patient or phantom. This factor increases with the area of the field irradiated and the thickness of the underlying tissues. In general, the increase is more marked for smaller field sizes and tails off with increasing field size.

**Related Article:** Backscatter


**Bad pixel**

*(Diagnostic Radiology)* Bad pixel is a term related to flat LCD monitors and large-area solid-state detectors in digital radiography. Bad pixels (or dead pixels) result from problems in the micro-electronic thin film technology (TFT) used in these monitors and detectors. Such pixels present either a black (or white) pixel on...
the monitor screen or the lack of detection (no response to input photons) from the detector. Each manufactured monitor or detector is generally tested to determine bad pixels. A discrete number of bad pixels can be corrected by software referring to a correction map and replacing them by values interpolated with surrounding pixels. Depending on the type of data readout, it is possible for a bad pixel to result in loss of information in a whole detector row or column.

**Baird-atomic system multi-crystal camera**

(_Nuclear Medicine_) The Baird-atomic system multi-crystal camera was an early scintillation camera developed in the 1970s to avoid image artefacts at high count rates and to get a better spatial resolution. It consisted of several small NaI(Tl) columnar crystals that were glued together to form a large detector block. The camera was then operated as an ordinary scintillation camera with Anger positioning logic. This system was primarily designed for high count rate scenarios and for clinical applications in cardiac studies of high activity bolus passage. A very high photon fluence rate would impinge on the crystal, causing pile-up effects both in the crystal from the scintillation light and also in the electronics. The former could then be reduced by altering the crystal geometry.

**Related Articles:** Scintillation camera, High count rates, High photon fluence rates, Cardiac studies

**Balanced FFE**

(_Magnetic Resonance_) See Fast imaging with steady state precession (FISP)

**Balanced gradients**

(_Magnetic Resonance_) A magnetic-field gradient with amplitude $G$ in direction $i$, varying as a function of time $t$, is balanced at the point $T$ in time if the zeroth moment of the gradient with respect to time is 0, i.e. if the following condition is fulfilled:

$$\int_{0}^{T} G_i(t) dt = 0$$

The condition for refocusing of phase dispersion induced by field gradients is fulfilled if this condition is fulfilled.

**Note:** After a 180° radio frequency (RF) pulse, a sign change of the gradient must be applied due to the inversion of phase induced by the RF pulse.

**Related Article:** Gradient motion rephasing (GMR)


**Ball bearing**

(_Diagnostic Radiology_) See Bearing

**Band gap energy**

(_General_) The band gap energy (or energy gap) is the energy range between the top of the valence band and the bottom of the conduction band. No electron states exist in this energy range. The width of the band gap energy determines the type of material – conductor, semiconductor, insulator.

For semiconductors, the band gap is a band with forbidden energy states, in which the electrons are forbidden to propagate. In order for an electron to get excited from the valence band to the conduction band, the electron must receive enough energy to transcend the band gap. The width of the band gap determines the conductivity of the semiconductor (Figure B.9).


**Band limiting**

(_General_) All electrical signals can be described in terms of their frequency content, and band limiting is the restricting or filtering of these signals in some way to a limited band of frequencies (Figure B.10).

The main purpose of band limiting is to preserve the desired information whilst minimising unwanted information (noise) where the two cannot be separated due to their different frequency content.

A second common use is to prevent unwanted signals from passing further through the chain of processes where their presence could inadvertently cause problems or indeed be processed in such a way that they transmute into frequencies within the bandwidth of the signal of interest.

Typical examples include lowpass, highpass, bandpass and bandstop filters, which are usually specified by the frequencies where the amplitude drops to 1/√2 or 3 dB of the passband amplitude.

One specific form of band limiting (lowpass) is required when analogue signals are to be digitised by an analogue to digital converter (ADC). Commonly referred to as an ‘anti-aliasing filter’, this filter prevents any signal greater than half the ADC sampling frequency from entering the conversion process, which could otherwise result in the higher frequencies aliasing and being transmuted into signals inseparable from the original data in the lower frequencies.

**Bandwidth**

(_Ultrasound_) Bandwidth in ultrasound applications is usually a measure of spectral width of a pulse. It is most commonly defined as the frequency band over which the amplitudes in the amplitude spectrum is over 70% (~3 dB bandwidth) of the maximum amplitude (or 50%, ~6 dB bandwidth). For a power spectrum, the power

![Figure B.9](image1)

**Figure B.9** Band gap.

![Figure B.10](image2)

**Figure B.10** Illustration of band limiting.

The condition for refocusing of phase dispersion induced by field gradients is fulfilled if this condition is fulfilled.

**Note:** After a 180° radio frequency (RF) pulse, a sign change of the gradient must be applied due to the inversion of phase induced by the RF pulse.

**Related Article:** Gradient motion rephasing (GMR)

Bar pattern

(Diagnostic Radiology) See Bar phantom

Bar phantom

(Diagnostic Radiology) Test devices consisting of parallel lines of an x-ray absorbing material separated by spaces of equal thickness to form line pairs (one line and the adjacent space). The width of a line and space is expressed in terms of line pairs per unit of length (lp/mm or lp/cm). This is in the spatial frequency domain. The test pattern is designed with sections containing different spatial frequencies as illustrated in Figure B.13.

Bar phantoms are used to evaluate imaging system blurring by determining the maximum spatial frequency at which the lines can still be ‘resolved’. The maximum frequency that can be resolved has a reciprocal relationship to the blur dimension.

Bar phantom

(Magnetic Resonance) Bar phantoms are used in MRI quality control to qualitatively assess the spatial resolution of the imaging system. Several groups of thin parallel plates are spaced in a test object. Each group of plates are placed a set distance apart from each other, which varies between each group. The distances used for the Eurospin test object are 2, 1.5, 1, 0.5 and 0.3 mm (Figure B.14).

To evaluate the spatial resolution, it is necessary to determine the smallest distance between plates that can be resolved. A conventional MR system should have a spatial resolution of 1 mm, but this can vary greatly with the pulse sequence and coil used as well as a number of other factors.

Bar phantom

(Nuclear Medicine) A bar phantom is a common name for physical and software phantoms used to evaluate the spatial resolution of an imaging system. In a typical bar phantom, the bar diameter and bar pitch is known and these parameters can be varied. The ability of the imaging system to separate and isolate the bars gives a measure...
of the systems performance in regards of spatial resolution. An example of a similar phantom (although not strictly a bar phantom) is the Jaszczak phantom as seen in Figure B.15.

### Barium (General)

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Ba</th>
</tr>
</thead>
<tbody>
<tr>
<td>Element category</td>
<td>Alkaline earth metal</td>
</tr>
<tr>
<td>Mass number A of stable isotopes</td>
<td>130 (0.106%), 132 (0.101%), 134 (2.417%), 135 (6.592%), 136 (7.854%), 137 (11.232%), and 138 (71.698%)</td>
</tr>
<tr>
<td>Atomic number Z</td>
<td>56</td>
</tr>
<tr>
<td>Atomic weight</td>
<td>137.33</td>
</tr>
<tr>
<td>Electronic configuration</td>
<td>1s² 2s² 2p⁶ 3s² 3p⁶ 3d¹⁰ 4s² 4p⁶ 5s² 4d¹⁰ 5p² 6s²</td>
</tr>
<tr>
<td>Melting point</td>
<td>1000K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>2170K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>3510 kg/m³</td>
</tr>
</tbody>
</table>

**History:** Barium was first identified by Carl Scheele in 1774 and extracted by Sir Humphry Davy in 1808. It is a soft alkaline earth metal whose name derives from the Greek word ‘bary’, meaning ‘heavy’. Barium metal reacts readily with oxygen in the air and with water. It is commonly found in nature as barite (barium sulphate) or witherite (barium carbonate).

**Medical Applications:** X-ray contrast agent – Barium has become the most widely used contrast agent for studies of the digestive tract due to its high density near room temperature and low toxicity. Typically, barium is administered to the patient orally as a ‘barium meal’, or through their rectum as a barium enema. Internally, the barium gradually coats the colon, improving image contrast due to its x-ray absorbing properties (barium is ‘opaque’ to diagnostic x-rays).

**Barium Fluoride Detectors:** Barium fluoride is an inorganic crystal with a very fast decay time (0.8 ns), sometimes used in time-of-flight PET detectors. Unfortunately, its photon yield is relatively small.

**Related Articles:** Contrast agent, Contrast enhancement, Contrast media, Detector PET, Time-of-flight techniques in PET

### Barium Fluoride (BaF)

**Nuclear Medicine** Barium fluoride (BaF) is an inactivated (not doped) inorganic scintillator. The scintillation decay can be described by two components; a slow (630 ns) and a fast component (0.6 ns). Together with the high atomic number, BaF is suitable for detector applications that require high detection efficiency per unit volume and a fast detector response.


### Barrier

**Nuclear Medicine** In nuclear medicine, a barrier can refer to both a physical tissue boundary defining a compartment and/or a physiological boundary preventing distribution of radiotracers.


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**Figure B.15** Two Jaszczak phantoms used for spatial resolution evaluation in small animal imaging equipment. The phantom is manufactured in such way that the hole width and pitch is known.
An example of a physical and physiological boundary is the blood–brain barrier (BBB). This restricts the passage of various chemical substances and bacteria into the brain whilst allowing passage for oxygen and vital nutrients.

Related Article: Compartment models

Barten model (Diagnostic Radiology) In medical imaging, it is important to have a visual consistency in how a given digital image appears on different monitors (CRT or LCD), taking into account also contrast sensitivity properties of human visual perception. National Electrical Manufacturers Association (NEMA) has prepared and kept under constant review a standard DICOM 3.14 (NEMA, 2011) to provide an objective, quantitative mechanism for mapping digital image values (pixel values or grey levels) into a given range of luminance of a medical imaging display (or of a light box). The relationship that the DICOM standard defines between pixel values and displayed luminance is based upon contrast sensitivity measurements and Barten model (Barten, 1999) of human visual perception. By this model, human contrast sensitivity is distinctly non-linear within a wide range of luminance. Barten’s model considers neural noise, lateral inhibition, photon noise, external noise, limited integration capability, the optical modulation transfer function, orientation and temporal filtering.

The human eye is relatively less sensitive in the dark areas of an image than it is in the bright areas of an image. This variation in sensitivity makes it much easier to see small relative changes in luminance in the bright areas of the image than in the dark areas of the image. Based on Barten model, a unit called just noticeable difference (JND) was defined. JND is the luminance difference of a given target under given viewing conditions that the average human observer can just perceive. The grayscale standard display function (GSDF) is the mathematically defined mapping of an input JND index to luminance values. GSDF of DICOM 3.14 covers the luminance range from 0.05 to 4000 cd/m² – this range corresponds to 1023 JNDs. The minimum luminance corresponds to the lowest practically useful luminance of cathode-ray-tube (CRT) monitors and the maximum exceeds the highest luminance of very bright light boxes used in mammography (Figure B.17).

Monitors that are calibrated to the DICOM 3.14 standard should always convert the same digital input values into the same luminance values. AAPM has provided the standard guidelines for the performance evaluation of electronic displays (AAPM, 2005). The AAPM document describes how to make quality assessments on medical display systems and also gives some acceptance criteria for the display systems.

Related Articles: Image display, Digital display, Brightness, Image perception


Hyperlinks: NEMA DICOM; AAPM

Base layer (Diagnostic Radiology) The base of a typical radiographic film, also known as film base (see the eponymous article).

Baseline (Ultrasound) In this context, the baseline refers to the horizontal line in a sonogram that defines zero velocity or frequency. This line can be shifted in most pulsed Doppler systems, thereby changing the useful velocity/frequency range in each direction. This may aid in reducing the affect of aliasing. If, for instance, the velocity range normally is displayed as −x to x cm/s (x dependent on the pulse repetition frequency), the baseline can be shifted to show velocities between 0 and 2x cm/s (Figures B.18 and B.19). This can be done when the direction of flow under investigation is known and the sonogram is uncontaminated by spectra from other vessels.

Baseline correction (Magnetic Resonance) In vivo NMR spectra frequently contain broad baseline features, due either to instrumental imperfections or to the presence of less mobile nuclei. This latter type of baseline is most significant in 31P NMR, where it originates from phospholipid molecules in cell membranes (Figure B.20). Removal of the baseline is frequently a prerequisite of peak assignment and peak area measurement, and this may be achieved either in the time domain or in the frequency domain.

Time Domain Methods: Apodisation – On the assumption that the baseline is due to short T2 components that decay rapidly in the FID, they may be suppressed by multiplying the time domain signal by a linear ramp that rises from zero at the beginning of signal
acquisition to unity after a few data points. This is a fairly crude technique that can result in undesirable line shape distortion.

Convolution Difference – Here, strong exponential line broadening of the time domain signal (see Spectral analysis) is applied to yield a spectrum in which all of the resonance peaks have been broadened to the extent that they are effectively eliminated from the spectrum and only the broad baseline remains. This is then subtracted from the original spectrum to achieve high-pass filtering.

**Frequency Domain Methods:** Curve fitting – Least squares fitting of one or more polynomial functions, or some other mathematical function, to the frequency domain spectrum is a popular means of baseline correction.

**Related Articles:** Peak areas, Peak assignment, Spectral analysis

**Bateman equation for secular equilibrium**

*(Nuclear Medicine)* The Bateman equation for secular equilibrium is a special case of the Bateman equation for parent–daughter relationship. The equation is applicable when the half-life of the parent is much longer than that of the daughter. One such parent daughter couple is $^{226}$Ra ($T_p = 1620$ years) and $^{222}$Rn ($T_d = 4.8$ days). In such a situation, $\lambda_p$ is close to zero and the original Bateman equation (see article Bateman equation) can be abbreviated to

$$A_d(t) = A_p(0)(1 - e^{-\lambda_d t}) \times BR$$

if the activity of the daughter is zero at $t = 0$. $A_d$ and $A_p$ are the activity of daughter and parent nuclei, respectively. $\lambda_d$ is the decay constant of the daughter nuclides. BR is the branching ratio for decay to the daughter when the parent has several decay paths available. After a time equivalent to 5 daughter half-lives, the activity of the daughter almost equals that of the parent nucleus ($A_d = 0.98A_p$), as seen in Figure B.21.

**FIGURE B.18** The sonogram range is from approximately ±58 cm/s. The peak systolic flows show aliasing with the peaks misrepresented.

**FIGURE B.19** The baseline has been changed, extending the velocity range in the direction of flow to 100 cm/s. This allows the sonogram to be displayed without aliasing and permits accurate measurement of peak velocities.

**FIGURE B.20** Phosphorus ($^{31}$P) NMR spectrum of the brain showing broad baseline feature.

**FIGURE B.21** The graph shows the build-up to a secular equilibrium. The half-life of the daughter is negligible compared to the half-life of the parent.
Bateman equation for transient equilibrium

**Bateman equation** The Bateman equation for transient equilibrium is a special case of the Bateman equation for parent–daughter relationship. The equation is applicable when the half-life of the parent is longer than the daughter’s half-life, but not long enough for \( \lambda_p \) to be approximated to zero (which will yield a secular equilibrium). One situation with a transient equilibrium is \(^{99}\text{Mo} \) (\( T_{1/2} = 66\) h) and \(^{99m}\text{Tc} \) (\( T_{1/2} = 6\) h). The original Bateman equation cannot be simplified unless \( \lambda_p = 0 \), so in the case of a transient equilibrium, the equation is

\[
A_d = \left[ A_p(0) \frac{\lambda_d}{\lambda_d - \lambda_p} \times (e^{-\lambda_p t} - e^{-\lambda_d t}) \right] \times BR + A_d(0)e^{-\lambda_d t}
\]

where

- \( A_d \) and \( A_p \) are the activities of the daughter and parent, respectively
- \( \lambda_d \) and \( \lambda_p \) are the decay constants for daughter and parent nuclides, respectively
- \( BR \) is the branching ratio for decay to the daughter when the parent has several decay paths available

Figure B.22 describes a hypothetical parent–daughter pair with \( T_p = 10 T_d \) and \( B.R. = 1 \).

When the ratio of the parent and daughter activities is constant – the parent and daughter are said to be in transient equilibrium. In some situations, it is interesting to know when the daughter activity has reached its peak, i.e. when \( dA/dt = 0 \). One example is the \(^{99}\text{Mo} \rightarrow ^{99m}\text{Tc} \) generator where \(^{99m}\text{Tc} \) is eluted. The maximum activity is attained if the generator is eluted every 23 h. The time at which the daughter activity reaches its maximum, \( t_{max} \), can be calculated using

\[
t_{max} = \frac{1.44 T_p T_d}{(T_p - T_d)} \ln \left( \frac{T_p}{T_d} \right)
\]

where \( T_p \) and \( T_d \) is the half-life of parent and daughter nuclides, respectively.

**Related Articles:** Transient equilibrium, Secular equilibrium, Bateman equation, Bateman equation for secular equilibrium

**Bateman equation in parent–daughter decay**

(Nuclear Medicine) The Bateman equations are used to describe parent–daughter relationships, i.e. situations where a radioactive parent nucleus decays to a radioactive daughter nucleus. The activity of the daughter nucleus is described by

\[
A_d = \left[ A_p(0) \frac{\lambda_d}{\lambda_d - \lambda_p} \times (e^{-\lambda_p t} - e^{-\lambda_d t}) \right] \times BR + A_d(0)e^{-\lambda_d t}
\]

where \( A_p(t), A_d(t) \) are the activities of the parent and the daughter, respectively, at time \( t \), and \( \lambda_p, \lambda_d \) are their respective decay constants. \( BR \) is the branching ratio for situations where there is more than one decay mode for the parent. The last in the equation is the residual daughter-product activity that might be present at time \( t \).

The Bateman equations can be applied to different situations, and the resulting equation depends on the relationship between the parent and daughter decay constants \( \lambda_p \) and \( \lambda_d \).

**Related Articles:** Transient equilibrium, Secular equilibrium, Bateman equation for secular equilibrium, Bateman equation for transient equilibrium

**Bayonet catch**

(General) Bayonet style of fitting of an electrical device (as a bulb) is used in some countries instead of screw fitting. The base of the device uses keyways (and not screw) to connect the device to the fixture base (bayonet mount). The device is locked in place by pushing it down and twisting to lock firmly.

**Beam alignment**

(Radiotherapy) The object of beam collimation is to produce a radiation beam of some required size and shape. In practice, the edge of the beam is not sharp as there is always a penumbra on the beam entrance surface due to leakage radiation from the collimator and also from geometrical factors such as the source dimension and the relative distance between the source and the collimator jaws. The magnitude of the penumbra increases with distance from the source and with the depth in phantom/patient. The usual statement of the beam size refers to the lateral distance between the 50% isodose lines at a reference depth in a phantom. An indication of the field size entering the patient is given by the field-defining light. Beam alignment is the procedure in which the field size defining light is made to coincide with the 50% isodose lines of the radiation beam projected on a plane perpendicular to the beam axis and at a standard SSD or SAD.

**Beam area**

(Ultrasound) The ultrasound beam area is defined as the area in a specified surface consisting of all points at which the pulse-pressure-squared integral, ppsi, is greater than a specified fraction of the maximum value of the ppsi in that surface. A common specified level is −6 dB. Units are metre squared, m\(^2\).

**Related Articles:** Beam width, Pulse-pressure-squared integral


**Beam arrangement**

(Radiotherapy) The beam and shielding device geometries are determined taking into account the 3D determination of the target position and the critical organ location. Generally, a single photon beam is of limited use in the treatment of deeply located tumours as it gives a higher dose at the depth of the maximum than at the tumour depth. Single fields are used for palliative treatments or for...
Beam attenuation

(Radiotherapy) Attenuation is defined as those processes by which a beam of x-ray or gamma radiation is reduced in intensity passing through some material. Since ionizing photon interaction does not always occur, the interaction processes are statistical in nature. The probability that a photon will interact with matter in one of several possible ways is a function of its energy and of the composition of the interacting material. It is generally assumed that the attenuation of gamma photons is exponential. When an absorber is placed in a gamma beam, the amount of radiations detected on the downstream side of the absorber depends on the absorber thickness, atomic number, density and the amount of scatter radiation that reaches the detector. The amount of scatter radiation that can reach the detector depends on the radiation beam cross section. When no scatter radiation reaches the detector, the geometry condition is called narrow beam or good geometry, whereas when a maximum amount of scattered radiation reaches the detector, the geometry condition is called broad beam or poor geometry condition.

In narrow beam geometry, a very small area of the absorber is irradiated and the intensity of radiation that reaches the detector is given by an exponential function of the following form:

\[ I(x) = I_0 e^{-\mu x} \]

where

- \( I(x) \) is the intensity of the transmitted beam at thickness \( x \)
- \( I_0 \) is the intensity of the incident beam
- \( x \) is the absorber thickness
- \( \mu \) is the linear attenuation coefficient

The intensity \( I \) is defined as the rate flow of radiant energy across a unit area. The attenuation coefficient is a probability per photon per unit path length that a photon interaction will occur.

In a broad beam geometry, the measured intensity is given by

\[ I(x) = B I_0 e^{-\mu x} \]

where

- \( B \) is the build-up factor
- \( I(x), I_0, x \) and \( \mu \) have the same meaning

The build-up factor \( B \) is introduced to correct for scattered radiation.

The linear attenuation coefficient is made up of three components (\( \tau \), \( \kappa \), \( \sigma \)) to take into account the main interaction processes of the photon interaction: \( \tau \) the photoelectric linear absorption coefficient, \( \kappa \) the pair production linear absorption coefficient and \( \sigma \) the Compton linear absorption coefficient.

The total linear attenuation coefficient \( \mu \) is therefore given by the sum

\[ \mu = \tau + \sigma + \kappa \]

The term ‘linear’ means that these coefficients are measured in units of inverse length, i.e. cm\(^{-1}\). Sometimes, it is useful to express distances in terms of the mass thickness, which is the product of the absorber density \( \rho \) and the thickness \( x \).

The beam attenuation can then be rewritten as

\[ I(x) = I_0 e^{-(\mu/\rho) x} \]

where \( \mu/\rho \) is called mass attenuation coefficient, compared with the linear attenuation coefficient that does not depend on the density of the absorber.

Sometimes it is convenient to utilise other coefficients that can be derived from the linear attenuation coefficient \( \mu \). The most common coefficients are indicated in Table B.1.

Beam collimator

(Radiotherapy) The treatment head of linear accelerators incorporate a fixed collimator and mobile jaws to achieve the radiation beam collimation. The photon beam collimation is achieved by a primary and a secondary collimator. The primary collimator defines the largest circular field size which is obtainable by the bremsstrahlung process of electrons accelerated into the linac accelerating tube and it consists in a conical opening into a tungsten block shaped on the transparent anode, which is used for the bremsstrahlung photon production. The photons, emerging from the primary collimator, struck a flattening filter to obtain the requested beam homogeneity. The secondary collimator consists of four movable blocks, two forming the upper secondary collimator and two the lower secondary collimator. The photon field size ranges from few millimetre squared to 40 cm × 40 cm at the linac isocentre distance. In modern linacs, independent jaw movements can provide asymmetric fields whose beam edge is coincident with the beam central axis. Recently, multileaf collimators (MLCs) have been introduced consisting of up to 120 collimating leaves (60 leaf pairs) whose movement motors are controlled by computer. The introduction of MLCs has permitted intensity-modulated fields in conformal radiotherapy either in the step-and-shot mode or in a continuous dynamic mode. The MLCs are covering field size up to 40 cm × 40 cm at the linac isocentre. A miniature version of the MLC (microMLC) has been also introduced, projecting 1.5–6 mm leaf width up to 10 cm × 10 cm at the isocentre distance. The microMLC is used in head and neck treatments and in special radiotherapy techniques.

### Table B.1

<table>
<thead>
<tr>
<th>Attenuation Coefficients</th>
<th>Linear</th>
<th>( \mu )</th>
<th>cm(^{-1})</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mass</td>
<td>( \mu/\rho )</td>
<td>cm(^2)/g</td>
<td></td>
</tr>
<tr>
<td>Atomic</td>
<td>(( \mu/\rho )(( A/N_a ))</td>
<td>cm(^2)/atom</td>
<td></td>
</tr>
<tr>
<td>Electronic</td>
<td>(( \mu/\rho )(( A/N_a/Z ))</td>
<td>cm(^2)/electron</td>
<td></td>
</tr>
</tbody>
</table>

\( \rho \) is the absorber density, \( A \) is the atomic weight, \( Z \) is the atomic number and \( N_a \) is Avogadro’s number (6.02 × 10\(^{23} \) atoms/gram atom).
as radiosurgery. In addition to primary and secondary collimators, the electron beam collimation is also obtained by adding different cone applicators. The thickness of the primary and secondary collimators usually corresponds to 3 ten value layers (TVL) in order to attenuate the average primary photon beam to less than 0.1% of its initial value. According to IEC recommendations, the maximum leakage should not exceed 0.2% of the open field value.

**Beam divergence**  
(Radiotherapy) Photon beams, produced using either a linear accelerator (LINAC) or an x-ray tube, are emitted in a cone pattern. Beam divergence refers to the increase in the circular cross-section of a photon beam with increasing distance from the source. The angle of divergence becomes more acute with increasing lateral distance from the central axis and as the SSD decreases. The maximum beam divergence is limited by the presence of a primary collimator in a conventional linear accelerator and by the anode angle in an x-ray tube:

\[
\frac{A}{B} = \left(\frac{f_a}{f_b}\right)^2
\]

When \(A\) and \(B\) are the cross-sectional areas at distance \(f_a\) and \(f_b\), respectively, the cross-sectional area is proportional to the square of the distance from the source (IAEA, 2005), as can be seen in Figure B.23.

**Abbreviation:** SSD = Source to surface distance.  
**Related Article:** Beam edge  

**Beam edge**  
(Radiotherapy) An ideal radiation field is when the cross-sectional area of the field is constant and non-zero inside the beam aperture and zero outside, in which case, the beam edge is defined by the field collimators. Practically, this cannot be achieved, as there is a physical penumbra at the beam edge due to a combination of geometrical penumbra, transmission through the collimators and scattered radiation from the irradiated volume. This penumbra can be defined as the lateral distance between the 80% and 20% values on a measured profile, relative to the central axis dose. The beam edge is defined as the 50% point on this penumbra and the field size is usually defined as the distance between the two 50% points on the dose profile (Figure B.24).

**Related Articles:** Beam divergence, Divergent beam edge, Secondary collimator, Dose profile, Penumbra  

**Beam energy**  
(Radiotherapy) It is essential that the energy of a beam is known since the response of most methods employed to measure the absorbed dose to water varies with energy. There are a number of different specifications that can be employed to indicate the energy of the x-ray beam produced by a piece of equipment.

The most obvious specification is the beam spectrum, but this is extremely difficult to measure directly or model using Monte Carlo. Therefore, alternative specifications have been used and these are different for different beams – kilovoltage x-ray photons, megavoltage x-ray photons and megavoltage electrons.

**Kilovoltage X-Ray Photons:** The criterion of choice for this range is HVL – half value layer, which is the thickness of an attenuating material required to reduced the air kerma rate in air to one half of its original value. HVL is typically quoted in millimetres of aluminium for beams up to 100kVp, and in millimetres of copper for kilovoltage beams greater than 100kVp. HVL is a reasonable guide to allow the user to assess tissue penetration and is used for the determination of properties in kilovoltage dosimetry protocols.

While the beams have a spectrum of energies (heterogeneous), it is also possible to calculate an ‘effective energy’ of a single energy (monoenergetic) beam that would produce the same HVL.

**Megavoltage X-Ray Photons:** At this higher energy range, the HVL specification does not vary significantly with energy; therefore, it is not suitable and other criteria have been developed.

One such qualifier is the nominal accelerating potential, which is simply the energy of the accelerated electron beam as it impacts the target in the head of the linear accelerator. This was used when dosimetry protocols were based on air kerma in air measurements,
but with the change to absorbed dose to water protocols, the penetrating power of the beam as it is attenuated by water or tissue has become more prevalent.

The most common attenuation-based specification is the quality index (QI or TPR 20/10) and is a special case of the tissue–phantom ratio (TPR) for depths of 10 and 20 cm. One of the main benefits of this is that it is independent of any electron contamination of the incident x-ray photon beam.

Other criteria that have been used tended to be related to the depth of maximum dose ($d_{max}$), but this makes them susceptible to any electron contamination at this relatively shallow depth.

Another specification is PDD(10), the percentage depth-dose value at 10 cm deep in water, which again raises the problem of electron contamination at the depth of maximum dose. It is possible to remove this by placing a 1 mm lead foil in the beam, and there are then additional corrections that must be made to take account of this.

Megavoltage Electrons: As with the x-ray photons, the beam of electrons that impacts the patient (or phantom) contains a spread (spectrum) of energies, and therefore the specification of choice has been the mean energy at the patient surface ($E_{m}$). This is derived from knowing the depth of 50% dose ($R_{50,0}$) and the following relationship:

$$E_{m} = CR_{50,0}$$

where

- $C$ is a constant (≈ 2.33 MeV/cm)
- $R_{50,0}$ is quoted in centimetres

The depth of 50% dose ($R_{50,0}$) is commonly used for the selection of stopping power ratios and reference depths ($z_{ref}$) in dosimetry protocols and is a similar idea to the penetrating power of the beam used in x-ray photon protocols.

**Abbreviations:** HVL = Half value layer, PDD = Percentage depth dose, QI = Quality index and TPR = Tissue–phantom ratio.

**Related Articles:** Calculation of absorbed dose, Mean electron energy, Tissue–phantom ratio, Beam quality, Quality index

**Beam flatness**

(Radiotherapy) Beam flatness is a measure of the homogeneity of the dose profile within a certain area of the central axis. The Institute of Physical Sciences in Medicine (IPSM) defines beam flatness as the ratio of the maximum dose within the beam by the minimum dose within the flattened area, i.e. MAX/MIN. The flattened area for a 40 cm field is the portion from the centre out to 3 cm from the edge of the 50% dose level, as shown in Figure B.25.

Similarly, beam symmetry is defined as the greatest value of equidistant points from the central axis within the flattened area, i.e. A1/A2 or B1/B2 from Figure B.24.

Linear accelerators (LINACs) will be designed to give a ‘flat’ beam at one particular depth, often 10 cm. To achieve this, flattening filters are placed in the way of the beam, to alter the photon distribution that exits the head. The photon beam produced by the interactions at the target is sharply forward peaked, which would produce significantly inhomogeneous dose profiles. The flattening filter is normally circularly symmetric and energy specific, as thicker filters will be needed for higher energy, and substantially reduces the dose rate at the centre. However, it also acts as a radiation filter and changes the energy spectrum across the profile, causing beam hardening in the centre of the beam as the low-energy photons are preferentially absorbed. Hence the penetrating power of the beam varies across the width, leading to unflatness in profiles at depths other than 10 cm.

If the flatness or symmetry of the beam is out of the tolerance values, then this may suggest the use of an incorrect filter, or an error in positioning. Alternatively, it may be that the monitor chambers are malfunctioning. The monitor chambers are multi-compartment ion chambers situated in the head of the LINAC to monitor the delivered dose. However, they are also a vital link in the feedback circuits that control the path of the beam using steering and bending magnets, in order to maintain the flatness and symmetry within tolerance.

Cobalt units tend to have flatter profiles than those produced by LINACs, but with wider penumbras due to the large physical source size.

**Abbreviations:** IPSM = Institute of physical sciences in medicine and LINAC = Linear accelerator.

**Related Articles:** Linear accelerator, Penumbra

**Further Reading:** Mayles, W. P. M. et al. 1988. Commissioning and quality assurance of linear accelerators, IPSM Report No. 54, Institute of Physical Sciences in Medicine, York, UK.

**Beam former**

(Diagnostic Radiology) The beam former is a metal filter used to produce an x-ray beam with specific hardness and intensity. It is used as a compensator while imaging objects with irregular shape. Typical examples of beam formers are beam restrictors and wedges used to image the heart (see the article on Beam restrictor) or the bow-tie filter used in computed tomography (see the eponymous article).

**Related Articles:** Beam restrictor, Bow-tie filter

**Beam hardening**

(Diagnostic Radiology) Beam hardening is the term used to refer to the increase in average x-ray beam energy when it passes through a material (Figure B.26). This occurs in a heterogeneous x-ray beam because the lower energies in the spectrum are...
absorbed preferentially. The flat and bow-tie filter present on computed tomography (CT) scanners will remove the lowest energy x-rays from the beam and reduce the effects of beam hardening.

In CT, when scanning objects with a circular or elliptical cross section, such as a body, the central rays pass through thicker parts of the object and therefore will undergo a greater amount of beam hardening than those at the periphery. The central channels in the detector bank will therefore ‘see’ apparently less attenuation than the outer ones, as a ‘harder’ beam undergoes fewer interactions (Figure B.27). The effect of this beam hardening can be seen as ‘cupping’ in the image (Figure B.28). This effect can be removed by applying calibration corrections to the attenuation profiles prior to reconstruction.

The extent of beam hardening is greater if highly attenuating materials such as bone or contrast agent are present in the path of the x-ray beam. This can lead to streaks and shading in the image (as the dark shadows in the middle of the CT scan in Figure B.29a). In these circumstances, the calibration corrections will not be sufficient to remove beam hardening and further software corrections, usually iterative bone corrections, are necessary to reduce these artefacts (Figure B.29).

Related Articles: Artefact, Bow-tie filter, Cone beam artefact, Helical artefact, Image artefact, Metal artefact, Motion artefact, Partial volume effect (artefact), Ring artefact

**Beam hardening** (Radiotherapy) The production of a high-energy photon beam is based on a Bremsstrahlung process in which a high-energy monoenergetic electron beam interacts with the nucleus of a material. The result of the interactions is a partial or complete loss of the electron energy and the consequent production of photons. The resulting photon Bremsstrahlung spectra may have energy up to the initial electron energy. If the Bremsstrahlung photon beam enters a material, a higher attenuation is obtained for the lower energy photons than for higher energy photons. This produces a change in the beam quality and the effect is called ‘beam hardening’. The energy degradation of the photon beam is also related, to a lesser extent, to the Compton scattering of the primary photons. In this case, the consequence is a beam softening. The net photon beam hardening alters the accuracy of the isodose distribution calculated by a treatment planning system, especially at large depths.

**Beam kernel** (Radiotherapy) The dose $D(x,y,d)$ at a given depth $d$ in a flat, homogeneous phantom calculated by the first principle, i.e. convolving the relative primary fluence distribution with the profile of the pencil beam distribution at the same depth is given by

$$D(x,y,d) = \int \int \Phi(a,b)K(x-a,y-b,d)da db$$

where

- $x$, $y$, $a$ and $b$ are lateral distance from the central axis
- $\Phi$ represents the relative fluence distribution for either an open or modified photon radiation field
- $K$ is the two-dimensional cross-sectional profile of the pencil beam dose distribution at the depth $d$
- $K$ is also called the convolution beam kernel and it is radially symmetric around the direction of the incident pencil

The pencil beam kernel $K$ incorporates the transport of scattered photons and secondary electrons in the phantom material. It is equal to the cross-sectional profile of a pencil beam at the specified depth. Pencil beam dose distributions in water or another tissue-equivalent material can easily be obtained with Monte Carlo calculation. The cylindrical geometry used for scoring the pencil beam dose distribution is shown in Figure B.30.

An 18MV pencil beam profile at a depth of 5 cm is shown in Figure B.31. The representation is for a kernel at a depth of 5 cm,
Beam limiting device

A beam limiting device consists of primary collimator, which defines the largest available circular field, and a secondary moveable collimator, which consists of four blocks (two forming the upper and two forming the lower jaws of the collimator) and which can provide rectangular or square fields. Modern linacs incorporate independent (asymmetric) jaws that can provide asymmetric fields.

Modern linacs are usually provided with multileaf collimators (MLC), which enable arbitrary shaping of the field. MLCs work either in static mode (step and shoot) or in dynamic mode (sliding window). MLCs are added to the secondary collimator or substitute the lower jaws of the secondary collimator according to the manufacturer. For very small fields, a micro-MLC can be used.

For electron beams, electron beam applicators are used to create a clinically useful beam.

Beam applicators are also used in superficial therapy (with low energy kilovoltage x-ray beams) and orthovoltage therapy (with medium energy kilovoltage x-ray beams).

Related Articles: Collimation, Multileaf collimator, MicroMLC, Electron applicators, Asymmetric jaws, Dynamic jaw collimation

Beam quality

(Radiotherapy) A beam limiting device is intended for limiting and shaping, or collimating the radiation field. All clinical radiation beams used in teletherapy must be collimated. Beam limiting device consists of primary collimator, which defines the largest available circular field, and a secondary moveable collimator, which consists of four blocks (two forming the upper and two forming the lower jaws of the collimator) and which can provide rectangular or square fields. Modern linacs incorporate independent (asymmetric) jaws that can provide asymmetric fields.

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Related Articles: Collimation, Multileaf collimator, MicroMLC, Electron applicators, Asymmetric jaws, Dynamic jaw collimation

X-Ray and Photon Beam Quality: In the case of a monoenergetic photon beam, the energy of the beam is the only index for the beam quality but in a Bremsstrahlung continuous spectrum, the half value layer (HVL) has been introduced as practical index for x-ray energies up to 400keV. The HVL is the thickness of a specified material that attenuates the x-ray beam to the half of its initial intensity. The measurement of HVL, based on beam attenuation, should be performed in a narrow beam geometry to avoid the scatter component reaching the detector. The HVL is given in aluminium thickness for low energy x-ray and in copper thickness for high x-ray energies. A better characterization of the x-ray beam quality could be given by the indication of the first and the second HVL. Another beam quality index of a heterogeneous photon beam is the effective energy, which is defined as the energy of a monoenergetic photon beam that has the same HVL as the heterogeneous x-ray beam. In accelerators, Bremsstrahlung x-ray beams are produced by accelerated high-energy electrons hitting a target. The photon beam quality is specified in megavolts (MV), i.e. the nominal accelerating potential of the highest energy photon in the continuous Bremsstrahlung spectrum produced by the electron beam of
Dosimetric protocols have also suggested the use of tissue–phantom distance (SSD) of 100 cm. The ratio 10/100 (Equations B.1 and B.2) is related to the photon attenuation in the exponential part of the depth-dose curve in water under reference conditions:

\[ I_{100} = I_0 \left[ \frac{(100 + 10)}{100 + 20} \right] e^{-10\text{MeV}/\text{HAT}}, \]  

(B.1)

where HAT is the half-attenuation thickness of water required to attenuate the photon beam of a factor of 2 in a broad-beam geometry:

\[ \frac{I_{10}}{I_{50}} = 1.19 e^{6.93/\text{HAT}} \]  

(B.2)

Electron Beam Quality: The beam quality of an electron beam differs from that in a photon beam since in an electron beam the variation of the physical parameters with depth in an absorber is diverse. The photon beam fluence decreases continuously with depth because of the attenuation while the electron fluence decreases at the same rate as the photon fluence only for depths where transient charged particle equilibrium (TCPE) is achieved without a significant variation of the energy spectrum of secondary electrons. The maximum energy and the mean energy of primary electron beam decrease continuously from a maximum at the phantom surface to zero at the depth of the maximum range in the material because of the continuous slowing down process. The electron beam inside the accelerator tube just before hitting the accelerator window is named intrinsic electron beam and exhibits a small energy and angular spread. As the intrinsic electron beam passes through the linac exit window, scattering foil, transmission monitor chamber and air reaching the phantom surface, the energy and angular spread of the beam increase significantly. In Figure B.33a and consecutive Figure B.33b, the electron energy spectra are shown, respectively, before the exit window (0), at the phantom surface (z) and at a depth z in the phantom (z). In the spectra, the mean energy, the most probable energy and the maximum energy of the beam are also shown.

In Figure B.34, an electron depth-dose curve is shown with the indication of 85% of the mean energy at the phantom surface, the mean energy 0 on the phantom surface and R50 the depth at which the absorbed dose falls to 50% of the maximum dose. The most probable energy 0 on the phantom surface is empirically related to the practical range 0 by 0 = 0.22 + 1.98 0 + 0.0025 2 where 0 is expressed in MeV and 0 in cm. The mean electron energy at the phantom surface is related to the half-value depth 0 by 0 = CR50 where 0.37 (cm) for water. 0 is calculated from the measured value of 0, the depth at which the ionisation curve falls to 50% of its maximum. 0 is determined by ionisation measurements in water by

\[ R_{50} = 1.029 I_{50} - 0.06 \text{ (cm)} \quad \text{ (for } 2 \leq I_{50} \leq 10 \text{ cm)} \]

\[ R_{50} = 1.059 I_{50} - 0.37 \text{ (cm)} \quad \text{ (for } I_{50} > 10 \text{ cm)} \]

The mean energy at a depth z in a water phantom is related to the practical range 0 by 0 = 0(1 − z / 0).

In Figure B.35, 85%, 50, and 0 are reported versus the most probable energy 0.

FIGURE B.33  Electron energy spectra.
Beam reproducibility

(Radiotherapy) Reproducibility of results of measurements indicates the closeness of the agreement between the results of the same measurements carried out under changed conditions. Reproducibility may be expressed quantitatively in terms of the dispersion characteristics of the results. Reproducibility is different from repeatability, which measures the success rate in successive experiments, possibly conducted by the same experimenters. Beam reproducibility is related to the variation of the output of photon and electron beams produced by a linear accelerator.


Beam restrictor

(Diagnostic Radiology) There are various beam restrictors used in x-ray imaging system. These are diaphragms, collimators, wedges, cones, etc. The purpose of the beam restrictor is to produce an x-ray beam with the size of the detector, thus avoiding exposure of the non-imaged parts of the body. Usually, these devices are made of highly absorbent metals like lead.

Diaphragm is usually only a metal piece (absorber) with an opening, mounted beneath the x-ray tube. It simply restricts the x-ray beam filed. Such circular diaphragm is used in dental x-ray tubes (usually restricting the beam to a circle of 6 cm at 20 cm from the focal spot). The simple tool (diaphragm) should not be mistaken with the light beam diaphragm in front of the x-ray tube housing, what is in fact a collimator.

A very simple beam restrictor is the cone. This is simply a metal extender (cylinder cone) attached to the diaphragm. Its length may vary – for example 20–40 cm. It has fixed aperture and is often used for radiographs of the skull, as it minimises the secondary radiation (which is normally prominent in radiography of bones).

More complex restrictor is the wedge filter (also called absorption filter or compensating filter), which is used for radiography of anatomical parts with significant absorption variation (chest radiography, skull radiography, etc.). For example, an aluminium wedge filter will produce x-ray beam with decreased intensity at one end (where the wedge is thicker). This beam can be useful in skull profile radiography, as it will produce a reasonable radiograph of the nasal bones (exposed with the decreased intensity beam), while the other part of the beam (exposed with the rest of the non-attenuated beam) will produce reasonable image of the other facial skull bones. A set of wedge filters are shown in Figure B.36.

Complex beam restrictors are collimators with specially shaped lead shutters (jaws) to produce x-ray beam. Most often these collimators produce rectangular fields (corresponding to the x-ray film sizes). When those collimators include light sources mimicking the x-ray beam shape, they are called light beam diaphragm (LBD). Some collimators are shaped to specific anatomical region (e.g. the nose) – Figure B.37. Collimators used in CT scanners restrict the beam width to the size of the slice thickness.

Some of these devices have a sensor that detects the size of the detector (film cassette) and automatically restricts the beam to this size. This is achieved by exact movement of the lead jaws of the

![Electron depth–dose curve.](image1.png)

**FIGURE B.34** Electron depth–dose curve.

![Variation of range parameters with the modal energy at the phantom surface.](image2.png)

**FIGURE B.35** Variation of range parameters with the modal energy at the phantom surface $E_{p,0}$.

![A set of special beam restrictors (wedge filters) used in DSA.](image3.png)

**FIGURE B.36** A set of special beam restrictors (wedge filters) used in DSA. (Courtesy of CGR.)
diaphragm (most often, the Bucky diaphragm). This system is also known as positive beam limitation device (PBL).

**Related Articles:** Attenuation, Step wedge, Filter compensating, Diaphragm, Collimator

**Beam spectrum**

(Diagnostic Radiology) An x-ray beam consists of photons of different energies. The spectrum is the range and distribution of photon energies for a specific beam as illustrated in Figure B.38.

The spectrum of an x-ray beam is determined and controlled by several factors including the tube anode material, type and amount of filtration in the beam and the ‘KV’ or electrical potential applied to the tube. The KV is the principle adjustable factor that is used to optimise the spectrum for specific applications.

**Beam symmetry**

(Radiotherapy) Beam symmetry is a measure of the homogeneity of the dose profile within a certain area of the central axis. The Institute of Physical Sciences in Medicine (IPSM) defines beam symmetry as the greatest value of equidistant points from the central axis within the flattened area, i.e. $A1/A2$ or $B1/B2$ from Figure B.25. The flattened area for a 40 cm field is the portion from the centre out to 3 cm from the edge of the 50% dose level, as shown in Figure B.25.

Similarly, beam flatness is defined as the ratio of the maximum dose within the beam by the minimum dose within the flattened area, i.e. MAX/MIN.

Linear accelerators (LINACs) will be designed to give a ‘flat’ beam at one particular depth, often 10 cm. To achieve this, flattening filters are placed in the way of the beam, to alter the photon distribution that exits the head. The photon beam produced by the interactions of accelerated electrons at the target is sharply forward peaked, which would produce significantly inhomogeneous dose profiles. The flattening filter is normally circularly symmetric and energy specific, as thicker filters will be needed for higher energy, and substantially reduces the dose rate at the centre. However, it also acts as a radiation filter and changes the energy spectrum across the profile, causing beam hardening in the centre of the beam as the low-energy photons are preferentially absorbed. Hence, the penetrating power of the beam varies across the width, leading to unflatness in profiles at depths other than 10 cm.

If the flatness or symmetry of the beam is out of the tolerance values, then this may suggest the use of an incorrect filter, or an error in positioning. Alternatively, it may be that the monitor chambers are malfunctioning. The monitor chambers are multi-compartment ion chambers situated in the head of the LINAC to monitor the delivered dose. However, they are also a vital link in the feedback circuits that control the path of the beam using steering and bending magnets, in order to maintain the flatness and symmetry within tolerance.

**Related Articles:** Linear accelerator, Penumbra

**Further Reading:** Mayles, W. P. M. et al. 1988. Commissioning and quality assurance of linear accelerators. IPSM Report No. 54, Institute of Physical Sciences in Medicine, York, UK.

**Beam weight**

(Radiotherapy) In case of a multiple field treatment, the beam weight is the contribution of a single beam to the treatment dose relative to the other fields. Treatment planning systems usually allow the choice of two weighting methods. For treatments at fixed SSDs, the weight represents a multiplying factor for the dose at a depth at the maximum dose, usually expressed as a percentage. For isocentric treatments, the field weight is defined as either the relative contribution of the beam to dose at $d_{max}$ or the relative contribution of the beam to the dose at isocentre. In this latter case, the beam weights can be altered to attain 100% at isocentre. This is done by applying the weights as a reciprocal of the number of beams. Weights can also be altered for applied weighted beams to achieve the required normalisation at isocentre.

**Beam width**

(Ultrasound) The width of an ultrasound beam is an important parameter affecting lateral resolution and resolution in the elevation plane. For circular transducers, it is symmetric about the beam axis but for many ultrasound devices including imaging systems using array transducers, it varies in different directions. The beam width is dependent on the distance from the transducer, focusing, gain settings, etc.

Beam width can be determined using a hydrophone. A hydrophone measures the pressure pulses and the pulse-pressure-squared integral (ppsi) is calculated for each point on a surface at the specified distance from the transducer or along a line in a specified direction, Figure B.39. The beam width is the distance between two points on this specified surface (line) in a specified direction passing through the point of maximum ppsi in that surface. It is defined...
as a fraction of the maximum ppsi value in the surface, normally −6 dB, which is one-fourth of the maximum.

**Related Article:** Pulse-pressure-squared integral

**Further Reading:** International Electrotechnical Commission, International standard, IEC 61157.

**Beamforming**

(*Ultrasound*) The principle of B-mode imaging is based on sequentially emitting short pulses and detecting the time of arrival and the amplitude of the echoes, Figure B.40. With a linear array transducer, hundreds of line measurements build up a cross-sectional image. These lines are updated continuously, allowing real-time imaging with a frame rate on the order of 10–60 images/s.

The beamformer is the part of the ultrasound scanner that determines and optimises the shape of the beams in both transmission and reception mode. As it is very difficult (impossible) to produce well-defined thin sound beams due to diffraction, etc., a number of different tricks are used to enable high-resolution imaging. In transmission mode for instance, the beamformer generates the time-delayed electric signals that drive the transducer elements, Figure B.41. Multiple focus zones, apodisation, dynamic focusing, dynamic aperture, delaying, summation, amplification, TGC and A/D-conversion are all processes that take part in optimised beamforming. The scanner computer steers and controls the main part of these processes.

When operating an ultrasound scanner, it’s most often possible to choose and change the position and number of transmission focuses. When more than one focus point is used, it is called choosing multiple focus zones. One or several indicators alongside the image marks at which depth(s) the transmitted beam is focused, Figure B.42. This position is set by the operator to the region of interest of the specific diagnostic examination. In case of multiple focus zones, the scan line information will contain information from several beam pulses, each produced with different focus depth, using different number of transducer elements and delay pattern, Figure B.43. This process increases the lateral resolution but reduces the frame rate.

**Related Articles:** B-mode, B-scanner, Apodisation, Dynamic focusing, Dynamic aperture, Time gain compensation (TGC)
Beam-on time

( RADIOTherapy) This is the time during which the beam is delivering radiation. Originally, treatment machines were controlled by timing devices, examples of which are Cobalt 60, orthovoltage and superficial machines where machine radiation output was measured in centigray (cGy) per minute (min), i.e. (cGy/min).

The beam-on time for linear accelerators (LINACs) and modern orthovoltage units is governed by a counter on which a number of units are set called monitor units. The monitor counts downwards from the number set to zero at which point the beam is switched off. Modern record-and-verify systems of LINACs count the monitor units up until the set limit is reached and the beam is switched off.

Typically, machine radiation output is measured in monitor units (MU) per 100cGy, i.e. (MU/100cGy or MU/Gy).

Beam's eye view

( RADIOTherapy) The technique of displaying the patient structures, which simulates the beam geometry in a plane perpendicular to the central axis is called beam's eye view (BEV). BEV displays the anatomy of the patient constructed from computed tomography (CT) image data as viewed by the point of view of the source of the radiation. The BEV display permits adjustment of the size and the orientation of the radiation beam to generate optimal beam arrangement to shield patient organ at risk. Divergence from the radiation source is taken into account so that the relationship of the beam collimation with respect to the patient anatomy is displayed correctly for the divergent beam. BEV could be used in planning conformal therapy where high gradient regions are customised to the 3D shape of the tumour to spare healthy tissues with small margins. BEV display is used not only in the determination of the best location and shape of the radiation fields but also in verifying the set up of the treatment isocentre.

Related Article: Organ at risk

Bearing

( Diagnostic Radiology) Inside rotating anode x-ray tubes, the anode and the rotor are within the glass envelope, supported by special ball bearings (the stator is fitted outside the glass envelope). The whole anode is supported by the bearing, which is of prime importance for the stable rotation of the anode. If the anode disc is very heavy, the single bearing would not be able to support it well (especially at high rotation speeds such as 9000 rpm) and the wobbly rotation will lead to an unstable focal spot and mechanical stress of the glass envelope.

Although the anode disc is connected to the rotor through a molybdenum stem (which is not a very good thermal conductor), the rotor and the bearings are heated during the x-ray exposure. If the ball bearings expand and block the smooth rotation, the target surface will immediately be overheated (even melted on places). Due to this reason, special dry-lubricated bearings are used (using metallic silver, lead, etc.). The preparation and lubrication of these bearings are often kept secret by the producers.

The high demands to the bearing of the anode (especially for the heavy high-power anodes) lead recently to a new design of the x-ray tube. This has been firstly applied by PHILIPS in their Super-Rotalix-Ceramic tube. The anode in this tube rotates on a stem, supported by bearings at each end. This mechanical support is only possible if the tube uses a metal envelope (with internal ceramic coating). This tube has three high voltage ceramic insulators (aluminium oxide) – for the two high voltage cables and the anode stem – see article on Ceramic x-ray tube with double bearings.

The latest developments in this field are x-ray tubes with special spiral groove bearings with liquid metal (Ga-In-Sn alloy) – see the articles on Liquid metal bearing.

Related Articles: Anode, Rotating anode, Glass envelope, Metal x-ray tube, Liquid metal bearing, Ceramic x-ray tube with double bearing

Becquerel

( Radiation Protection) The Becquerel (Bq) is the SI unit of radioactive decay. It is the quantity of radioactive material in which one atom is transformed per second:

$$1 \text{ Bq} = 1 \text{ transformation per second}$$
The unit was named after the French physicist, Henri Becquerel, who is credited as having discovered the phenomenon of radioactivity in 1896.

Prior to the adoption of Becquerel as the SI unit of radioactivity, the unit was the Curie (Ci), named after Pierre Curie. This unit was equivalent to the rate of disintegration of 1 g of radium-226, which was \(3.7 \times 10^{10}\) disintegrations per second. Thus, the equivalence between Bq and Ci is

\[1 \text{Ci} = 3.7 \times 10^{10} \text{ Bq}\]

Related Article: Curie


BED (biological effective dose)

(Radiotherapy) See Biological effective dose (BED)

BEIR

(Radiation Protection) See Biological effect of ionising radiation

Bending magnet

(Radiotherapy) In many high-energy LINACs, the electron beam from the waveguide is horizontal, and must be rotated 90° in order to direct it at the target.

The simplest system rotates the beam through 90°, as seen in Figure B.44, and was the design for early LINACs. This design has the advantage that it requires very little vertical height. Unfortunately, any transverse, angular or energy displacement of the electrons entering this system results in a positional displacement for the output electrons, resulting in a broad focal spot. This design is no longer used commercially, in favour of the following designs.

The 270° magnet bends the beam back on itself, crossing the incident beam, as shown in Figure B.45. One method for achieving this is the ‘locally tilted pole gap’, which has adjustable entrant, and exit pole faces, enabling optimal beam focusing. Siemens uses this method.

An alternative method for the 270° design is the ‘three sector uniform gap’ magnet, which is used by Varian, and consists of three uniform-field dipole magnets with interconnecting drift tubes.

The slalom magnet or 112.5° magnet rotates as shown in Figure B.46, giving a small focal spot and requiring less vertical room in the treatment head. This design is favoured by Elekta (previously Philips).

Bernoulli effect

(Ultrasound) Bernoulli’s principle states that for an incompressible fluid, if no work is performed on the fluid and there are no frictional or other energy losses, then for steady flow, there is conservation of energy along a streamline as follows:

\[\frac{\rho V^2}{2} + \rho gh + P = \text{constant}\]

where

\(\rho\) is the fluid density
\(V\) is the velocity at a point in the streamline
\(P\) is the pressure at the point
\(h\) is the height of the point above a datum
\(g\) is the gravity

The three components of energy are the dynamic energy, pressure energy and gravitational potential energy.

In the human circulation, the principle and equation is useful in describing the change in energy as velocity increases, for example in a stenosis. If there is negligible change in gravitational energy, for example in a stenosis across a short distance, then the equation simplifies to

\[\frac{\rho V^2}{2} + P = \text{constant}\]

By comparing pre-stenotic velocities with those in the stenosis, an estimate of the pressure reduction in the stenosis can be made.

BERT

(Radiation Protection) See Background equivalent radiation time (BERT)

Bessel function

(Nuclear Medicine) Bessel functions are solutions to the Bessel differential equation
$$x^2 \frac{d^2 y}{dx^2} + x \frac{dy}{dx} + y (x^2 - \alpha^2) = 0$$

and where $\alpha$ is the order of the function. Bessel functions are used to describe electromagnetic waves, heat conduction and problems related to lattices. In nuclear medicine, Bessel functions are used in image processing, for example image filtering.

**Beta decay**

*(Nuclear Medicine)* The process in which an atom decays by beta-particle emission is called beta decay. It can occur in two different ways: $\beta^-$ and $\beta^+$ decay.

The kinetic energy of beta particles has a continuous spectrum, which depends on parent and daughter nuclear states participating in the decay.

In a $\beta^+$ emission, a neutron transforms into a proton, an electron and a neutrino. The $\beta^+$ decay is of no greater interest in clinical tomographic imaging since the range of the electron in tissue is only on the order of a few millimetres; thus, for a $\beta^+$ particle originating inside the human body, the probability of reaching a detector is low. $\beta^+$-emitting nuclei are used in preclinical imaging and for some therapeutic purposes.

$\beta^+$ emission, on the other hand, is one of the basic conditions for PET imaging. The different steps in a $\beta^+$-decay are schematically represented in Figure B.47. $\beta^+$ decay is essentially a transformation of a neutron into a proton, a positron (positively charged electron) and a neutrino:

$$p^+ \rightarrow n + e^+ + \nu + \text{energy}$$

Emitted positrons will travel a few millimetres in tissue before they are brought to a near stop. The distance travelled depends on the positron kinetic energy. When stopped, the positron combines with an electron and for a short while they form a short-lived particle known as positronium. Both particles are consumed in a process called annihilation, in which two photons are created. These two photons can be registered using a PET imaging system. The photon energy is equivalent to the mass of each particle (511 keV) and the photons are emitted in opposing directions (conservation of momentum for a stationary electron–positron pair). However, the particles are seldom brought to a full stop, and the residual momentum results in a deviation from a 180° emission angle. The positron range and deviating emission angle degrades the spatial resolution in PET imaging (separate article).

A schematic representation of a $\beta^+$-decay. A $^{11}\text{C}$-atom decays, which produces a neutrino, a $^{11}\text{B}$ atom and a $\beta^+$-particle. The $\beta^+$-particle is slowed via interactions with surrounding matter. When brought to a near or total halt, the positron forms a particle, a so-called positronium, with an electron. The positronium is consumed in an annihilation process that emits two photons with 180° opposing directions.

**Related Articles:** Annihilation PET, Spatial resolution PET


**Beta decay**

*(Radiation Protection)* Beta decay (sometimes called beta minus decay) is one of the mechanisms of radioactive decay. In the nucleus, a neutron is transformed into a proton with the emission of an electron, which, for historical reasons, is called a beta particle: $n \rightarrow p + e^-$. An anti-neutrino ($\nu^-$) is also emitted. The transformed (daughter) nucleus has one less neutron and one additional proton. Gamma rays may also be emitted such that the daughter nucleus decays to its ground state.

An example of beta decay with no emission of gamma rays, as it decays directly to the ground state, is the decay of phosphorous-32 into stable sulphur-32:

$$\frac{32}{15}P \rightarrow \frac{32}{16}S + \beta^- + \nu^*$$

The decay scheme is shown in Figure B.48.

The beta particle (an electron, also sometimes called a negatron) emitted can have a kinetic energy of any value up to a definite maximum energy, which is characteristic of the nuclide involved. The shape of the spectrum is generally similar for many nuclides – see Figure B.48. Where more complex spectra are observed, these are found to be composed of several simple spectra superimposed.

The average energy of the beta particles in the spectrum is approximately one-third of the maximum energy. The anti-neutrino carries off the difference in energy between the beta particle and the maximum energy in beta minus decay. The neutrino is often omitted from decay equations.

**Related Articles:** Beta particles, Decay schemes, Neutrinos, Nucleus, Radioactive decay, Stable nuclei

**Beta particle**

*(General)* The term used for an electron when it is emitted from an atom as a result of one type of radioactive decay – beta decay. Symbol is $\beta^-$. 

**FIGURE B.47** Scheme of different steps in a $\beta^+$-decay.

**FIGURE B.48** The decay of $^{32}\text{P}$. Energy of the atom is plotted on the $x$-axis and the atomic number on the $y$-axis.
A beta particle is emitted from an unstable nucleus (radionuclide) as the result of the transformation of a neutron into a proton and an electron: $n \rightarrow p^+ + e^-$. The electron (beta particle) is ejected from the nucleus with kinetic energy of any value up to a definite maximum energy, which is characteristic of the decay process (Figures B.48 and B.49).

The use of the term beta particle is historical as three different types of emissions from radioactive decay (alpha, beta and gamma) were identified in the early days of studying radioactive decay by Ernest Rutherford (1871–1937) and others, but their exact nature – type, charge and mass – was not known.

**Related Articles:** Radioactive decay, Radioactivity

### Beta radiation

*(Radiation Protection)* Beta radiation is a specific form of ionising radiation arising from the emission of beta particles from radioactive material where the radionuclide involved is decaying by beta decay.

Beta particles are electrons emitted during beta decay.

**Related Articles:** Beta particles, Beta decay, Electron, Ionising radiation, Radionuclide

### Beta+ radiation

*(Radiation Protection)* Beta+ radiation (beta plus radiation, $\beta^+$) is composed of beta+ particles, which are commonly called positrons (positive electrons), and is emitted from radionuclides undergoing radioactive decay through the decay process normally referred to as ‘positron emission’.

When the beta+ particle, or positron, leaves the nucleus, it loses its kinetic energy by interacting with surrounding atoms. When it has lost its kinetic energy, it is annihilated by combining with an electron from one of the surrounding atoms to form two photons, referred to as annihilation radiation:

\[
\beta^+ + e^- \rightarrow \gamma + \gamma
\]

The mass of the particles is converted into electromagnetic radiation according to Einstein’s equation, $E = mc^2$. The two photons each have an energy of 0.51 MeV and they travel, to a first approximation, in opposite directions so that their net momentum is zero.

**Related Articles:** Beta radiation, Beta decay, Radioactive decay

### Betatron

*(Radiotherapy)* The betatron was a device that accelerated electrons and was developed for research purposes in 1940, but its application for treatment in radiotherapy was soon recognised.

Electrons are accelerated in an evacuated doughnut-shaped ring by a magnetic field generated by an alternating current of between 50 and 200 Hz (see Figure B.50).

Electrons are injected from a filament and held in a central orbit by the magnetic field. The increasing magnetic field induces an accelerating voltage in the direction of the electron ring similar to the secondary coil of a transformer. The electrons can then be deflected using electrodes placed in the evacuated doughnut. In the 1950s, these machines played an important part in radiotherapy, but their role has been taken over by linear accelerators. This is partly due to the greater output possible with LINACs (10 Gy/min for linacs versus 1 Gy/min for betatrons). Although betatrons can produce electrons with energies from a few MeV up to about 45 MeV, the clinically advantageous sharp cut-off seen with electrons is lost after about 20 MeV; so in therapy, the useful range of energies is from about 4 to 20 MeV and linacs can easily produce this electron energy range.

**Abbreviations:** LINAC = Linear accelerator.


### Bethe–Bloch equation

*(Radiation Protection)* The Bethe–Bloch equation describes the energy lost by interactions of fast-moving charged particles (e.g. protons, alpha particles, but not electrons/beta particles) traversing matter. It is named after two theoretical physicists: Hans Albrecht Bethe and Felix Bloch.

Charged particles moving through matter interact with the electric fields of the electrons and nuclei of atoms in the material. The interaction excites or ionises the atoms. This implies the transfer of energy from the incident particle.

The equation describes the energy loss per unit distance travelled (otherwise known as the stopping power of the material traversed) – it describes the absorption process of charged particles travelling through matter, and implies that the particles will have a finite range in matter whereby all incident and secondary particles have been totally absorbed within the medium (as opposed to x- or gamma radiation that is attenuated exponentially with at least some transmission through the material and out the other side).

The Bethe–Bloch equation, together with the Klein–Nishina differential cross-section, Möllöe scattering theory and others, attempts to describe interactions between ionising radiation and matter at an atomic level, and forms the mathematical basis for radiation dosimetry based on Monte Carlo statistical modelling.
Related Articles: Stopping power, Collision mass stopping power, Linear stopping power, Klein–Nishina differential cross section, Molière scattering theory

Biangular anode disc
(Diagnostic Radiology) X-ray tubes with biangular anode discs are used for dual focus tubes, where there is a need of very small effective focal spot. Such tubes are used for cardio, neurological and other x-ray examinations requiring radiographs with high spatial resolution. Biangular anode construction is predominantly used for powerful rotational anode x-ray tubes.

In these tubes, the line-focus principle allows for reducing the effective focal spot, by using target with smaller bevel. From Figure B.51, it is obvious that the same filament wire will produce smaller effective focal spot at angle α, compared with those at angle β.

The separation of the effective focal spots creates two actual focal spots (Figure B.51), which allows better heat distribution when consecutive exposures with different focal spots are made (compared with the situation of overlapping filament – see Focal spot, actual).

Such x-ray tubes could have a third smaller angle (closer to the anode stem), thus producing even smaller focal spots (microfocus x-ray tubes used in macroradiography).

Related Articles: Stationary anode, Rotating anode, Target, Line-focus principle, Biangular anode disc, Focal spot actual, Focal spot effective, Focal spot

Bias
(General) Bias is an electronics term referring to the establishment of a background DC voltage or current at a point in a circuit. A bias is designed into a circuit in order to keep a device or circuit element operating around a preferred quiescent voltage or current. For example, in an amplifier design, a transistor may be biased to keep it operating in a linear region of its characteristic. In the simple amplifier design shown later, resistors R1 and R2 act as a potential divider to keep the base of the transistor at a predetermined bias voltage. The input AC signal coupled to the transistor base via a capacitor rides on this bias voltage – see the electric circuit.
The higher the atomic number of the atom the greater the binding energy of the electrons in a particular shell, varying for the K-shell as the square of the atomic number. The energy of the electrons in the outer shells is complicated by the interactions between electrons.

**Related Article:** Ionisation

### Binomial excitation

*(Magnetic Resonance)* Binomial excitation is a technique using composite RF excitation pulses to achieve frequency selectivity for use in, for example fat suppression or in magnetisation transfer. In binomial excitation, a train of RF pulses is applied, with each pulse having a relative flip angle determined by the binomial coefficients $q_{n,m}$:

$$q_{n,m} = \binom{n}{m} = \frac{n!}{(n-m)!m!}$$

When binomial coefficients are used to relatively weight a series of RF pulses equally spaced part by a time $\tau$, it can be shown that the resultant RF excitation spectrum has null points at frequencies at odd multiples of $1/2\tau$ Hz from the centre frequency. Frequencies ‘off resonance’ from the centre frequency by $1/2\tau$ Hz will then experience no net excitation.

As an example, consider the binomial series (1, 3, 3, 1). For a total flip angle of 90°, the binomial excitation will consist of the four RF pulses: $\pi/16$, $3\pi/16$, $3\pi/16$ and $\pi/16$. In Figure B.52, note that for an ‘on resonance’ spin, the accumulation of flips in this case results in the required total of 90°. For an off-resonant spin with a frequency $1/2\tau$ different from the resonant frequency, the transverse component slips through 180° between each RF application. As a result, the accumulation of flips in this case amounts to a net flip of zero.

Equally, it can be shown that where every second binomial coefficient is negative, there is a null at the ‘on resonance’ centre frequency of the RF spectrum and peaks at frequencies $1/2\tau$ Hz either side of the centre frequency. Appropriate application of a binomial excitation can then excite either resonant or off-resonant tissue. For example, as fat and water have slightly different precession frequencies, the fat being off-resonance from water by about 220 Hz in a 1.5 T system, either fat or water can be selectively excited by this technique through an appropriate choice of $\tau$.

**Binomials** *(General)* In algebra, binomial refers to polynomials with two terms, or the sum of two monomials.

**Bioeffects** *(Radiation Protection)* Ionising radiation causes biological damage at a cellular level. Such radiation exposure may lead to direct interactions on the DNA within the nucleus of each cell, or by more indirect means – interactions with the water and other biological molecules within the cytoplasm, leading to chemical reactions and thus to biological damage. For more detail on these interactions, see *Radiation damage*.

From an understanding of the effects of ionising radiation at the cellular level, it is possible to relate the resulting cellular damage to observable biological effects at a macroscopic level – i.e. the effects to the organism (human or animal) exposed. Firstly, though, it is necessary to classify such effects:

The biological effects of ionising radiation on an exposed individual are classified as being either somatic, i.e. effects that occur within the lifetime of the individual irradiated, or genetic, which are effects occurring in progeny due to radiation damage to the reproductive cells of the parent. Such genetic (or hereditary) effects may only become evident after several generations.

A further classification describes those effects that manifest in individuals who were exposed to ionising radiation in utero, such damage being called teratogenic effects. These effects may be either somatic to the foetus exposed, manifesting in utero or postpartum, or heritable (genetic) in nature, affecting future progeny as the individual exposed in utero reaches adulthood and has children of his or her own.

Somatic effects may also be further classified by the probability of occurrence in irradiated populations:

Deterministic effects are those definite effects that will occur in all persons exposed to high doses of radiation – the ‘radiation syndromes’ or ‘radiation sickness’.

Stochastic effects are those that occur only to a proportion of those exposed based purely on chance. Cancer is the main stochastic effect. If a population is exposed to ionising radiation, then a number of people will get cancer. The more the radiation dose received, the larger the number of people expected to get cancer.

All damage expressed in either the person exposed, or progeny, may also be referred to as adverse effects or adverse radiation effects.

**Related Articles:** Radiation damage, Repair of radiation damage, Adverse effects, Adverse radiation effects, Stochastic effects, Deterministic effects

**Biological dosimeter** *(Radiation Protection)* Biological dosimetry means the detection, and if possible, the quantification, of exposure to ionising radiation, using biological indicators. There are, in fact, some methods to measure the biological effects of the ionising...
radiations in the human body. The bio-dosimetry assessment is based on the simple assumption that the exposure or contamination will produce effects that are measurable (given a certain variability due to various responses). The most sophisticated example is the measurement of chromosome aberrations. This kind of analysis has been carried out from 1982 and is reliable and sophisticated; due to its complexity, it can be applied only in selected cases. The principle is that ionising radiation causes structural chromosome aberrations, a proportion of which give rise to chromosome fragments. When cells divide, some of these fragments form small nuclei within the cytoplasm. These micronuclei can be counted, providing an in situ biological dosimeter. In the United States, the Federal Office for Radiation Protection is the most competent body in the field of chromosome aberrations. Competent institutions are present in each European country. Other indicators for the exposure to ionising radiation are the lymphocyte lifetime, the presence of erythemas of different levels, etc.

**Biological effect of ionising radiation (BEIR)**

*(Radiation Protection)* Several major international committees and several national scientific bodies, relevant to radiation protection, came into existence in the mid-1950s. Namely, the United Nations Scientific Committee on the Effects of Atomic Radiation (UNSCEAR), the Committee on the Biological Effects of Atomic Radiation (BEAR).

The BEAR was renamed the Committee on the Biological Effects of Ionising Radiation (BEIR) in 1972, set up by the US National Academy of Sciences. The UNSCEAR and the BEIR committees have continued their work up to the present, periodically reviewing the levels of radiation to which human populations are exposed and improving assessment of the somatic and genetic risks of radiation exposure.

The BEIR reports include the following:

- **BEIR Committee.** (The Advisory Committee on the Biological Effects of Ionising Radiation). 1972. BEIR-I: The effects on populations of exposure to low levels of ionising radiation. Division of Medical Sciences, the National Academy of Sciences, National Research Council, Washington, DC. 2006. (Generally known as the BEIR Report.)

- **BEIR Committee.** 1979. BEIR-II: The effects on populations of exposure to low levels of ionising radiation. Draft Report. Division of Medical Sciences, Assembly of Life Sciences, National Research Council, National Academy of Sciences. (Generally known as the BEIR-II Report.)

- **BEIR Committee.** 1980. BEIR-III: The effects on populations of exposure to low levels of ionising radiation. Final Report. Division of Medical Sciences, Assembly of Life Sciences, National Research Council, National Academy of Sciences. (Generally known as the BEIR-III Report, Final.)


**Biological effective dose (BED)**

*(Radiotherapies)* The physical radiation dose received by a tissue or tumour does not necessarily reflect the resulting biological effect. This will also depend on the radiosensitivity of the tumour or tissue and the fractionation scheme. The biological effective dose (BED), also known as the extrapolated response dose (ERD), is a convenient tool that allows quantification of the biological effect and is derived from the linear-quadratic model, as shown in Figure B.53. It allows different fractionation schedules to be compared in terms of their biological effectiveness and for alternative regimes to be calculated, for example to correct for an unwanted interruption of treatment.

In Equation B.6, the BED is equal to the log cell kill by radiation. However, this ignores the effect of repopulation. For irradiated tissue that is concurrently repopulating, we must subtract the log cell kill from the log cell kill by irradiation. For Equation B.7, where

\[
BED = nd \left[ 1 + \frac{d}{\alpha/\beta} \right] - K(T - T_{delay})
\]

where

- $T$ is the overall treatment time
- $T_{delay}$ is the delay time (from beginning of treatment) before the onset of significant repopulation
- $K$ is the biological dose per day required to compensate for ongoing tumour cell repopulation

The biological effective dose received by a uniformly irradiated tissue that is concurrently repopulating.

Equation B.7 is valid when treatment is delivered with well-spaced fractions, but when two or more fractions are delivered per day, there may be increased biological damage as a result of the incomplete
The biological effective dose equation requires the addition of a factor, $h$, when multiple fractions per day are delivered to reflect the increased damage resulting from the incomplete repair of sublethal damage.

The conventional practice to use the $\alpha/\beta$ value used in the BED calculation as a subscript to both the BED and its numerical symbol emphasises the point that the biological effective dose is not a real physical dose. For example, a stated BED of 120 Gy indicates that an $\alpha/\beta$ of 10 Gy was used to calculate that particular biological dose. Biological doses expressed in Gy can be added to other Gy values to provide a measure of resultant effect, but it is not permissible to add Gy values to Gy values.

Since BED is derived from the linear-quadratic (LQ) model, the same limits apply regarding its validity (for more details, see the article on LQ model). It should also be noted that a single BED calculation may not be sufficient: where there are significant dose inhomogeneities, the BED will vary. ‘Hot spots’ of doses higher than that prescribed may occur within the target volume and involve critical normal tissues. These should be considered in any calculations. For example, the BED in the region of a 110% ‘hot spot’ will require the dose per fraction, $d$, to be multiplied by 1.10 in the calculation. Potential overdosing problems can be prevented by performing any BED matching for the calculation of biological equivalent dose fractionation schedules at the ‘hot spot’ region, but it should be noted that this may result in a reduction in tumour control, particularly if inhomogeneities extend to ‘cold spots’ within the tumour volume. In the future, the use of biological effective dose-volume histograms as suggested by Wheldon et al. (1998) or the concept of equivalent uniform dose (EUD) as suggested by Niemierko (1997) may prove of benefit when assessing non-uniform planning volumes. Other dose-volume models for tumour control probability and normal tissue control probability are beginning to be incorporated into some computerised treatment planning systems where they may be useful for comparing plans but do not as yet provide a reliable model for absolute calculations.

The parameters $\alpha/\beta$ and $K$ in the BED equations are both tissue specific; therefore, appropriate values must be selected. Since precise values of $\alpha/\beta$ are rarely known for individual patients, it is generally recommended that a range of values are used in multiple calculations. In practice, generic values of 5 and 10 Gy for tumour and 3 Gy for late normal tissue effects are frequently used, the latter often reduced to 2 or even 1.5 Gy in the case of CNS, which is known to be particularly sensitive to the fractionation regime. Precise values of $K$ and $T_{delay}$ are also unknown. For most late-responding normal tissues, the proliferation rate is usually so small that $K$ can be neglected, i.e. $K = 0$. Dale et al. (2002) provide some guidance on selection of $K$ and $T_{delay}$ for various human tumours, but it should be noted that these values have large uncertainties associated with them.

Values of BED can be difficult to interpret because they are not physical dose. Most clinical experience of the tolerance of normal tissues has been obtained from treatment regimes that use 2 Gy fractions. Therefore, ‘equivalence’ can be simply obtained and expressed as the equivalent dose in 2 Gy fractions (EQD2). Figure B.54. Note that EQD2 is sometimes called the normalised total dose in 2 Gy fractions (NTD):

A worked example of the use of BED to find an alternative fractionation scheme that produces the same biological effect to the tumour is given in Figure B.55. The dose per fraction required to deliver the same biological effect to the tumour in 20 fractions as the regime that delivers 60 Gy in 30 fractions is calculated. In this example, repopulation effects are ignored and generic values of $\alpha/\beta$ of 10 Gy for the tumour and 3 Gy for normal tissue are used. The effect of this alternative regime on the normal tissues can be evaluated by calculating the equivalent dose in 2 Gy fractions. The clinician can draw on their clinical experience of tolerance doses to determine if this regime will be acceptable in terms of the probability of inducing an adverse effect. It is important to note that it is not possible to maintain the BED both for tumour and for normal tissues. As expected from a hypofractionated regime, the effect of maintaining the same biological effect to the tumour increases the biological effect to normal tissues.

More details on the use of BED with many worked examples can be found in the book by Dale and Jones (2007).

In the United Kingdom, the Royal College of Radiologists (2008) in their latest guidance on dealing with unplanned interruptions of radiotherapy recommend that radiobiological calculations are performed by suitably trained physicists or clinicians, preferably who have attended an appropriate training course. It cannot be emphasised enough that quantitative radiobiological assessments inform but do not replace clinical judgement.
**Abbreviations:** BED = Biological effective dose, ERD = Extrapolated response dose, EQD2 = Equivalent dose in 2 Gy fractions and NTD = Normalised total dose in 2 Gy fractions.

**Related Articles:** Adverse effect, Alpha-beta ratio, Dosevolume histogram, Fraction, Fractionation, Interruption of treatment, Linear-quadratic (LQ) model, Normal tissue complication probability, Repopulation, Radiosensitivity, Tolerance, Tumour control probability


**Biological half-life**

*(Nuclear Medicine)* The disappearance with time of a pharmaceutical in a biological system can be described by a law similar to that of radioactive decay. The biological half-life is therefore defined as the time required for half of the pharmaceutical to disappear from the biological system. The biochemical elimination of this substance from the body is achieved through processes such as excretion, metabolism and diffusion.


**Biological parameter**

*(Radiotherapy)* In radiotherapy, a biological parameter is a term in a mathematical model used to describe a radiobiological phenomenon. An example of such a parameter is the alpha–beta ratio in the linear-quadratic model that describes the response of mammalian cells to radiation.

**Related Articles:** Alpha–beta ratio, Linear-quadratic (LQ) model, Normal tissue complication probability, Radiobiological models, Tumour control probability

**Biological purity**

*(Nuclear Medicine)* Biological purity may be defined as the absence of undesirable biological components in a radiopharmaceutical, for example free from microorganisms and toxic microbial by-products, such as bacterial endotoxins. Almost all radiopharmaceuticals are intended for parenteral administration, i.e. given intravenously, subcutaneously or muscularly, and must be prepared by aseptic processing.

Measurement of the sterility, microbiological control and bacterial endotoxin testing (BET) must be performed regularly and on randomly selected batches of the radiopharmaceutical to check the sufficiency of the aseptic technique, or whenever there might be a need for it.

All sterile radiopharmaceuticals should contain no pathogenic or nonpathogenic living organisms, meaning that all products that are intended for parenteral administration must be sterilised by autoclaving or by membrane filtration. The most common method is to filter the solution through sterile 0.22μm Millipore filters.

All products designed for intravenous administration are required to be free from pyrogens. These are metabolic products of microorganisms that cause a pyretic response (fever) in the patient between 45 and 90 min post injection. Endotoxin is the most significant pyrogen and detected by BET. Endotoxin is not removed by membrane filtrations, which is why a solution may be sterile but not pyrogen free. Since the most likely sources of pyrogens are impure water and chemicals used in the preparation, pyrogenic contamination can be prevented by using high-quality water and chemicals, and glassware dry-heated at 250°C for 3 min.


**Biological response models**

*(Radiotherapy)* Biological response models are mathematical models that describe radiobiological phenomena. In general, biological response models describe how the probability or frequency of a specific biological response changes with some other parameter, usually dose level. Examples of such models include the linear-quadratic model, tumour control probability and normal tissue complication probability.

**Related Articles:** Linear-quadratic (LQ) model, Normal tissue complication probability, Radiobiological models, Tumour control probability

**Biological target volume (BTV)**

*(Radiotherapy)* The gross tumour volume (GTV), clinical target volume (CTV) and planning target volume (PTV) are well-established volumes used in radiotherapy planning. The biological target volume (BTV) is another planning volume that allows for inhomogeneous dose prescriptions.

For example, the tumour cell density may not be homogeneous throughout an organ, or even within a tumour, within an organ. If there is a region within the PTV that is known to be particularly aggressive or prolific, then this region can be further delineated to become the BTV and receive a higher dose. This information is now becoming more readily available with the complementary use of modern imaging techniques, including MR spectroscopy, SPECT and PET. This approach is called ‘Biological Image Guided Radiotherapy’.

**Abbreviation:** BTV = Biological target volume.

**Related Articles:** Gross tumour volume (GTV), Planning target volume (PTV), Clinical target volume (CTV)

Biplane cine system

(Diagnostic Radiology) Some fluoroscopic systems need to present simultaneous images of an organ from two angles. This is very important for cardio examinations (invasive radiology). These systems use two C-arms, each with a separate x-ray tube and image intensifier, connected to a separate cine camera. Contemporary biplane systems may use fast digital image recording instead of cine camera. The movement of both C-arms around the patient should be independent to allow various angles of observation, as well as space for the surgical intervention. It is possible that the movement of the C-arms is isocentric, thus always keeping the heart in the middle of the observation field. Naturally, the biplane fluoroscopic system has two TV monitors – one for each viewing angle.

Related Articles: Fluoroscopy, Image intensifier, C-arm

Biplane imaging

(Diagnostic Radiology) There are various x-ray procedures that benefit from simultaneous imaging in two directions or planes. Two-directional or biplane imaging specifically overcomes some of the limitations of a one-directional view through a body, which does not provide depth information.

A major application of biplane imaging is for the heart. With the two fluoroscopic images, the physician can view the coronary arteries from two directions as they are being filled with contrast media. Defects such as a stenosis, which are visible in one plane, might not be visible in the other plane because of the many different directions and orientations of the arteries.

Biplane (and sometimes multi-plane) imaging is an advantage for fluoroscopic (or radiographic) imaging. Other imaging modalities such as CT and MRI present the same effect by reconstructing images in multiple planes.

Related Article: Biplane cine system

Bipolar gradient

(Magnetic Resonance) A bipolar gradient is composed of two gradient pulses that are equal in magnitude and duration but opposite in sign. This re-phases stationary spins while flowing spins experience a phase shift (Figure B.56). The velocity \( v \) in the direction of the gradient corresponds to a net shift in the phase image. The main application is in flow quantification and in phase-contrast angiography. By putting two bipolar gradients of opposite sign together (back-to-back), the phase of the uniformly flowing spins can be refocused. In this way, the signal loss due to flow-induced dephasing is minimised in gradient-echo sequences.

Related Article: Phase contrast angiography


Bipolar pulse in amplifier for radiation detector

(Radiation Protection) The linear amplifier for radiation detector is used to amplify and shape a linear tail pulse (voltage or current) (Figure B.57). The amplitude of a pulse (signal) may be used to evaluate the energy of radiation recorded by a detector.

If monopolar pulses (Figure B.58a) are used, their amplitude is measured relative to a baseline (zero). The shift of the baseline causes an error in estimation of a pulse amplitude, thus in radiation energy. The bipolar pulse (Figure B.58b) has almost equal amounts of positive and negative area. Thus, the net voltage is zero and a bipolar pulse can pass by a capacitor without alteration of the baseline. The signal-to-noise ratio (SNR) is poorer after bipolar shaping than after monopolar, but the pulse width is the same. The monopolar shaping is used at low counting rates, whereas bipolar shaping at high counting rates.


Birdcage coil

(Magnetic Resonance) A birdcage coil is a particular design for an RF volume coil that can be used for both RF transmission and reception in MR imaging. The basic structure of a birdcage coil consists of a number of parallel conductors distributed at equal spacing around the surface of a cylinder and orientated along the direction of the static magnetic field. Birdcage coils provide the high uniformity required of volume coils, providing good constancy in gain throughout an imaged volume when used in receive mode and by the principle of reciprocity, good uniformity in flip angle in the imaged volume when used in transmit mode (Figure B.59).

Theoretical analysis shows that an ideal cylindrical volume coil with a sinusoidal distribution of current with radial angular position \( \phi \) on the coil surface demonstrates excellent uniformity of excitation throughout a volume in transmit mode (and by reciprocity, excellent uniformity in receive mode). The distribution of current in the conductors of a birdcage coil can be shown to approximate this ideal.

Bismuth germanate (BGO)

(Nuclear Medicine) Bismuth germanate (BGO) is an alternative to the common NaI(Tl) crystal. It has a high atomic number (83) and high density (7.13 g/cm\(^3\)) and is therefore suited for the detection of high-energy photons such as those from positron emitters. It is also less fragile and hygroscopic than NaI(Tl), allowing a wider use in more demanding conditions. One of the disadvantages of BGO is the relatively low light yield, which is only 10%–20% that of NaI(Tl). Another disadvantage is the high refractive index (2.15), which makes the transition of light photons between the crystal and photomultiplier tube less effective.

BGO has a relatively short signal decay time compared to NaI(Tl) and some other scintillators.

Bit

(General) Bit (binary digit) is the basic unit of information storage in a computer system. The bit can take two values – 0 or 1. A byte is a collection of 8 bits, representing $2^8 = 256$ values (from 0 to 255). File sizes are usually expressed in kilobytes ($1\, kB = 10^3$ bytes), megabytes ($1\, MB = 10^6$ bytes), etc.

Bit depth

(Diagnostic Radiology) See Matrix size

Blackening

(Diagnostic Radiology) See Film blackening

**Related Articles:** Inorganic scintillators, Scintillators, NaI(Tl) detector crystal

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**Blank exposure**

(Diagnostic Radiology) A blank exposure is a radiographic film that was not exposed because of an equipment malfunction, operator error or other abnormal condition.

**Bloch equations**

(Magnetic Resonance) The Bloch equations are differential equations describing the behaviour of a single spin magnetisation vector in a magnetic field. The basic equation is given by

$$\frac{d\mathbf{M}}{dt} = \gamma \cdot (\mathbf{M} \times \mathbf{B}) - \left( \frac{M_x}{T_2} \mathbf{i} + \frac{M_y}{T_2} \mathbf{j} \right)$$

where

- The vectors $\mathbf{M}$ and $\mathbf{B}$ are the magnetisation vectors and applied magnetic field respectively
- $\gamma$ is the gyromagnetic ratio
- $M_0$ is the spin magnetisation equilibrium value
- $M_x$, $M_y$ and $M_z$ are the components of $\mathbf{M}$ along the directions $\mathbf{i}$, $\mathbf{j}$, and $\mathbf{k}$

The first term reflects the interaction of the $\mathbf{M}$ and $\mathbf{B}$ vectors, and represents simple precession where $\mathbf{B}$ is the static main field $B_0$.

The second term represents the observed decay of the transverse components of magnetisation $M_x\,\mathbf{i}$ and $M_y\,\mathbf{j}$ according to $T_2$ relaxation, and the recovery of the longitudinal component of magnetisation $M_z\,\mathbf{k}$ to $M_0$ according to $T_1$ relaxation.

The equation can be solved to determine the time course of the magnetisation from some initial magnetisation $\mathbf{M}$ in response to an applied $\mathbf{B}$ field consisting of an RF field $B_1(t)$ and the static field $B_0$.

**Block design**

(Magnetic Resonance) During a functional MRI (fMRI) study, the subject undertakes a series of tasks known as a paradigm. These
Block design

In the simplest type of AB block design, an epoch of stimulus (an ‘on-period’, A) is followed by an epoch of rest (an ‘off-period’, B). Typically, they may each be 30 s long, with a 5 min scan consisting of five blocks and five intervals. As the stimulus and rest epochs are distributed evenly throughout the scan, the stimulus and rest signals should be equally affected by temporal variations in scanner sensitivity, subject concentration and subject motion (Figure B.60).

Statistically, block designs are very powerful. By carefully choosing appropriate block lengths, it can be ensured that the subject’s haemodynamic response function maximises during the on-period, and returns completely to baseline during the off-period. For this reason, the block design often produces strongly significant results.

The basic design can be elaborated by introducing additional blocks of stimuli (e.g. ABC block design), or randomising the order of the different blocks (ABBCACB… instead of ABCABCABC.…).

Related Articles: fMRI (functional magnetic resonance imaging), Haemodynamic response function, Event-related design


Block tray

A block tray is attached to the treatment head at a suitable distance from the machine in order to take the attenuation of the tray into account. See Table B.3 for examples of tray attenuation factors (Metcalfe, 1997, p. 233).

Blocks may be placed on the tray if it is flat, i.e. gantry at 0° or 180°. When oblique or lateral beams are blocked, the blocks are attached to the tray by means of screws. The screw holes are drilled through the tray and into the block in the case of Perspex.

### Table B.2

<table>
<thead>
<tr>
<th>Thickness and Type of Block</th>
<th>Transmission (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>10 cm lead</td>
<td>0.7</td>
</tr>
<tr>
<td>10 cm low-melting-point alloy</td>
<td>3.8</td>
</tr>
</tbody>
</table>

TABLE B.3
Blocking Tray Attenuation Factors

<table>
<thead>
<tr>
<th>Type of Blocking Tray</th>
<th>Attenuation Factor</th>
</tr>
</thead>
<tbody>
<tr>
<td>Low-density metal (honeycomb)</td>
<td>0.995</td>
</tr>
<tr>
<td>Perspex (6 mm)</td>
<td>0.970</td>
</tr>
<tr>
<td>Perspex (10 mm)</td>
<td>0.951</td>
</tr>
</tbody>
</table>

Low-density metal (of the kind used to construct aircraft, such as the alloy Dural) can also be used. These trays are thin and easy to work with and are made with many holes drilled into them.

Related Article: Block transmission factor


Blocking layer
(Diagnostic Radiology) Blocking layer is a common term referring to a layer of material included in a radiation detector or image receptor to block exposure from an unwanted type of radiation.

Blood–brain barrier
(Magnetic Resonance) The blood–brain barrier (BBB) is a cellular structure that acts primarily to protect the brain from certain chemicals in the blood, while still allowing essential metabolic function. The barrier is composed of very tightly packed endothelial cells in brain capillaries. This higher density alters the permeability of brain capillaries, so that some substances are prohibited from entering the brain tissue, while other substances are permitted to enter freely.

Blood–brain barrier leakage
(Magnetic Resonance) An intact BBB is a prerequisite for using a common gadolinium-chelate MRI contrast agent as an intravascular tracer for the assessment of cerebral perfusion in dynamic susceptibility contrast MRI (DSC-MRI). When the BBB is damaged, the contrast agent will leak into the extravascular extracellular space. This may lead to reduced $T_1^*$ as well as $T_1$ relaxation times of the extravascular spin population. These two effects will counteract the signal loss normally caused by the contrast-agent-induced susceptibility effects in DSC-MRI, leading to errors in, primarily, the quantification of cerebral blood volume (CBV).

Two ways of minimising the effects of BBB leakage in DSC-MRI have been presented. One approach is to correct for the effects using algorithms in the post-processing step, as proposed by, for example Weisskoff et al. (1994), Haselhorst et al. (2000), and Vonken et al. (2000). The second approach is to give a small pre-dose of contrast agent before the actual DSC-MRI experiment (Sorensen and Reimer, 2000). This pre-dose will lead to a shortening of $T_1$, and when the main bolus arrives, the $T_1$ relaxation is more or less saturated and no further shortening of $T_1$ will occur.

Related Articles: Blood–brain barrier, Dynamic susceptibility contrast MRI, Cerebral blood volume


Blood oxygenation level–dependent (BOLD) contrast
(Magnetic Resonance) The brain requires a steady flow of oxygen to function, which is transported by haemoglobin in the blood. In functional MRI, blood oxygenation level–dependent (BOLD) contrast is the most commonly used mechanism for detecting brain activity.

Whilst it has been shown that a BOLD response directly reflects an increase in neural activity (see Logothetis et al., 2001), the understanding of its neural origins and evolution remains unclear. At a basic level, the process can be traced as shown in Figure B.61:

1. An fMRI task or stimulus causes an increase in neuronal activity in brain areas related to the task.
2. The increase in neuronal activity induces complex neurovascular coupling.
3. The neurovascular coupling activates a haemodynamic response (see related article: Haemodynamic response function) such that blood flow to the active brain region is increased and the local level of oxyhaemoglobin becomes elevated relative to the level of deoxyhaemoglobin.
4. As oxyhaemoglobin is diamagnetic, and deoxyhaemoglobin is paramagnetic, the increase in oxyhaemoglobin prompts a local increase in $T_2^*$. Thus, the active brain region is marked by increased MR signal on $T_2^*$ weighted images (typically, the signal intensity rises by a few percent).

In fMRI, BOLD responses are detected by statistical analysis of a time series of images. The intensity of each voxel within the image is typically statistically compared to a model of the expected BOLD

![Figure B.61](Image 126x73 to 486x107) The basis of fMRI BOLD contrast. (Image courtesy of Arthurs and Boniface.)
response to determine whether activation of a particular brain region has occurred.

As the magnitude of the BOLD response varies with field strength, fMRI sensitivity can often be improved by moving to higher field strengths.

**Related Articles:** fMRI (functional magnetic resonance imaging), Oxymoglobin, Haemodynamic response function


**Blurring**
(Diagnostic Radiology) See Detail resolution

**B-mode**
(Ultrasound) Brightness-mode (B-mode) imaging is the most common way to display diagnostic ultrasound measurements. Emitting short pulses and detecting the time of arrival and the amplitude of the echoes build the image. In B-mode imaging, the echo amplitudes are converted to gray levels where white dots correspond to high amplitudes. With a linear array transducer, hundreds of line measurements build up a cross-sectional image. These lines are updated continuously, allowing real-time imaging with a frame rate on the order of 10–60 images/s (depending on penetration depth, number of lines, number of focus zones, etc.) (Figures B.62 and B.63).

**Related Articles:** A-mode, M-mode, B-scanner

**BMUS**
(Ultrasound) The British Medical Ultrasound Society is a multidisciplinary society organisation with a membership of sonographers, physicians, technologists, scientists, physicists and engineers. The society’s stated aims are to advance the science and application of ultrasound in medicine and to provide education, guidance and information for its members and the public. It achieves this through its annual meetings, journal and website.

**Hyperlink:** BMUS: [www.bmus.org](http://www.bmus.org)

**Body coil**
(Magnetic Resonance) The body coil is an RF coil built into the MRI system. It surrounds the bore of the magnet and is hidden from view by the protective covers lining the bore. The body coil may be used both as a transmit coil and a receive coil (Figures B.64 and B.65). Body coils are volume coil designs and display good uniformity over a large field of view. Where a large, uniform field of view is required, the body coil will frequently be utilised as RF transmitter and receiver. For example, the body coil is particularly suitable for general imaging of the thorax and abdomen.

However, where a higher SNR is required and a smaller field of view is adequate, it will be generally more appropriate to take advantage of the better SNR achievable using a surface coil or a smaller volume coil (e.g. a head coil) as receiver. In addition, the

**FIGURE B.63** B-mode imaging.

**FIGURE B.64** MRI system with the covers removed, showing the body coil (innermost elements) and the surrounding gradient coils.
use of separate receiver coils can enable the implementation of parallel imaging techniques. The body coil is generally used as the transmit coil even with other coils being used as the receiver. As a transmitter, the body coil has the advantage of delivering a uniform distribution of flip angles. The only exception to utilisation of the body coil as a transmitter is where the imaging coil used also has transmit capability (e.g. some head and extremity coil designs).

**Body contour**

*(Radiotherapy)* In general, the body contour represents the outline of the patient as indicated on an image to be used for treatment planning, for example CT or MRI scan slices. Other, simpler, traditional methods include the use of lead wire or mechanical rods to form the shape of the patient surface. These are then traced on paper before being digitised onto the planning system.

The body contour also indicates the shape of the skin surface onto which the treatment field is incident. Typically, this will produce an oblique angle of incidence and so some distortion of the dose distribution will result.

**Abbreviations:** CT = Computed tomography and MRI = Magnetic resonance imaging.

**Body habitus**

*(General)* A term used in medicine to describe the physical characteristics of an individual in relation to body size. For example, ‘poor body habitus’ in an ultrasound report would describe an individual who was obese.

**Body mass index (BMI)**

*(General)* This is a statistical measure of the weight of a person scaled according to height. Body mass index is defined as the individual’s body weight divided by the square of their height.

**Body protein monitor**

*(Radiotherapy)*

**Background:** The body comprises distinct and measurable body compartments. The status and rate of change of these compartments reflects the health of a person and the response of treatment for a specific disease. Whereas measurement of body weight is a basic and useful parameter, it can be a misleading measure of response to treatment, which may increase oedema and fat while depleting protein.

An important compartment is the total body protein (TBP), which is a measure of muscle and visceral mass. It is determined directly by measurement of the total body nitrogen (TBN). The relationship between nitrogen and protein is

\[ \text{TBP} = 6.25 \times \text{TBN} \]

TBN is regarded as the superior measure of protein status in diseased subjects.

**Methods:** In vivo methods are used to obtain TBN values before and after therapy, so as to determine the effect of therapy.

The measurement of normal values is required so as to compare patient values, according to a defined nitrogen index, \( \text{NI} = \frac{\text{TBN}_p}{\text{TBN}_n} \), where \( p \) designates the patient and \( n \) the normal values for healthy subjects.

The importance of this index relates to the impact on disease prognosis, and has clinical significance in cancer and renal disease.

In vivo interrogation techniques are used. These can give a measure of the whole compartment, or in some cases, the spatial distribution of the compartment.

Nitrogen is measured in a body protein monitor (BPM) using a CF252 or PuBe neutron source. The patient is moved over a collimated neutron beam. The fast neutrons emitted by these radioactive sources are moderated in tissue and captured by nitrogen and hydrogen nuclei in the patient. The 11.4 MeV ground-state gamma ray emitted after nitrogen capture can be measured by NaI detectors, being of higher energy than all background radiations. The hydrogen gamma ray (2.2 MeV) is very intense and easily detected. The ratio of nitrogen to hydrogen yields eliminates the dependence of gamma ray attenuation on body habitus.

**Abbreviations:** BPM = Body protein monitor, NI = Nitrogen index, TBP = Total body protein and TBN = Total body nitrogen.

**Related Articles:** In vivo body composition, Total body nitrogen, Total body potassium, Total body water, Total body fat

**BOLD contrast**

*(Magnetic Resonance)* See *Blood oxygenation level–dependent (BOLD) contrast*

**Boltzmann distribution**

*(Magnetic Resonance)* Classically, when a strong external magnetic field \( B_0 \) is applied to nuclei with non-zero nuclear spin, these nuclei are forced to precess about an axis either parallel or antiparallel to the magnetic field. Spins precessing parallel to \( B_0 \) are in a state of lower energy than those precessing antiparallel to it, such that the parallel spin state (lower energy state) claims the greatest share of the spin population. The population difference varies with temperature according to the Boltzmann distribution:

\[
\frac{N_{Ap}}{N_P} = \exp \left( \frac{\delta E}{k_B T} \right)
\]

where

- \( N_{Ap} \) is the number of spins in the antiparallel state
- \( N_P \) is the number of spins in the parallel state
- \( T \) is the temperature
- \( \delta E \) is the energy difference between the two states
- \( k_B \) is the Boltzmann constant

When a patient (body temperature \( \sim 310 \text{ K} \)) is placed in a 1.5 T scanner, the population difference between the two levels is approximately 1 in \( 10^6 \). Such a population difference means that the vector sum of spins will be non-zero and will point parallel to the magnetic field. In other words, in the presence of a magnetic field \( B_0 \), tissue becomes magnetised with net magnetisation \( M_0 \). This induced magnetisation \( M_0 \) is the source of signal in all MR experiments.
Bolus (Radiotherapy) Bolus is a tissue equivalent material placed directly on the skin surface to even out the irregular patient contour and thereby provide a flat surface for normal beam incidence.

In principle, the use of bolus is straightforward and practical; however, it suffers a serious drawback: for megavoltage photon beams, it results in the loss of the skin sparing effect in the skin under the bolus layer (i.e. skin sparing occurs in the bolus). Bolus is placed in contact with the skin to achieve one or both of the following:

1. **Increase the surface dose:** To increase the surface dose, a layer of uniform thickness bolus is often used (0.5–1.5 cm), since it does not significantly change the shape of the isodose curves at depth. Several materials have been developed commercially for this purpose, which are floppy and malleable and easily shaped to the patient's surface. However, wet towels or gauze wrapped in cellophane offer a low-cost substitute.

2. **Compensate for missing tissue:** To compensate for missing tissue or a sloping surface, a custom-made bolus arrangement can be made that conforms to the patient's skin on one side and yields a flat perpendicular incidence to the beam (see Figure B.66). The result is an isodose distribution that is identical to that produced on a flat phantom; however, skin sparing is not maintained. This can be overcome by retracting the bolus (taking divergence into account) as in Figure B.66.

A common material used for this kind of bolus is wax, which is essentially tissue equivalent and when heated is malleable and can be fitted precisely to the patient's contour. Bolus can also be used to compensate for lack of scatter, such as near the extremities or the head during total body irradiation (TBI). Saline or rice bags can be used as bolus in these treatments.

**Related Articles:** Compensating wedge, Tissue compensation, Compensating filters, Compensation, Compensator

### Bolus injection

(Nuclear Medicine) When discussing a bolus injection in nuclear medicine, one is referring to a prompt intravenous injection of a small volume of the radiopharmaceutical. An example of an examination where it is very important that a bolus is injected is first-pass radionuclide ventriculography.

**Bolus of tracer** (Magnetic Resonance) A bolus of a tracer is the single administration into a blood vessel over relatively short time to achieve a blood concentration at a relatively high level. In the theoretical modelling of tracer kinetics, the concept of an ‘ideal’ bolus is introduced. The injection or input of such an ‘ideal bolus’ is assumed to be instantaneous, i.e. the dose distribution over time is described by the Kronecker delta function. Tracers employed in medical imaging are often radiolabelled substances or contrast agents (for computerised tomography or magnetic resonance imaging).

**Related Articles:** Dynamic susceptibility contrast MRI, Bolus tracking

### Bolus tracking

(Magnetic Resonance) When performing contrast-enhanced MRA (see Contrast-enhanced angiography), a contrast media are injected intravenously, and images are acquired as the bolus reaches the area of interest. The timing of the MRA acquisition is essential to obtain optimal image quality for the vascular type (arterial or venous phase) and/or the vascular territory (large or small vessels). Therefore, the determination of the time delay between the injection of contrast agent and the arrival of this agent bolus to the region of interest is important. There are two main methods to determine accurate timing of the data acquisition from ‘tracking the contrast agent bolus’.

The first method utilises a test bolus, i.e. a small amount of contrast agent. The test bolus is injected, and a special bolus imaging scan is started. This is a dynamic scan of usually one slice with good time resolution to track the bolus. As dynamic scans are acquired during the passage of the test bolus, reviewing the image set gives the optimal time point after contrast injection to start the data acquisition in the subsequent CE-MRA scan. This method requires two injections, and a revision of images in between scans. Furthermore, it assumes that the timing of the bolus does not change, for example due changes in the heart rate.

The second method uses a real-time view of the area of interest. A simplified version of the CE-MRA sequence is run dynamically, and the operator can interactively start the subsequent CE-MRA scan as the contrast is seen to reach the vessels to be studied. This method requires knowledge of the profile order of the subsequent CE-MRA scan – if the order is centric, the scan should be started as the contrast reaches the area of interest, while it, in the case of linear order, should be started before the contrast arrival, to obtain the k = 0 samples with correct timing.

**Related Articles:** Contrast-enhanced angiography, Contrast media

### Bone

(General)

<table>
<thead>
<tr>
<th>Density</th>
<th>1000–2000 kg/m³</th>
</tr>
</thead>
<tbody>
<tr>
<td>CT number</td>
<td>400–3000 HU</td>
</tr>
</tbody>
</table>

Bone is a rigid organ that forms the endoskeleton of vertebrates. There are 206 bones in the adult human body. Bone has multiple functions, including to provide shape, movement and protection of the organs of the body, to allow sound transduction, to produce blood cells, to store minerals, growth factors and fat, to detoxify and to achieve acid–base balance. Bone consists predominantly of mineralised osseous tissue containing calcium hydroxyapatite, which provides bone with its rigidity and strength as well as relatively low density due to its pseudo-honeycomb internal structure. There are two types of
Bone, including cortical (or compact) bone and trabecular (or cancellous) bone. Cortical bone makes up 80% of bone by mass and provides a hard smooth outer layer. Trabecular bone forms a porous interior network of bone, providing its lightweight structure and a high surface area, containing bone marrow and blood vessels. Bone consists of osteoblast cells, which build bone and produce hormones; osteocyte cells, which form and maintain bone and function in calcium homeostasis; and osteoclast cells, which resorb and remodel bone.

Bone is assessed in medicine by various techniques, most commonly conventional radiography or computed tomography (CT) to assess anatomical structure. Functional imaging can be carried out, such as a nuclear medicine bone scan by administration of a radioisotope (technetium-99m) labelled phosphate analogue tracer, methylene diphenosphonate (MDP). Osteoporosis is usually assessed by a DEXA (dual-energy x-ray absorptiometry) scan to calculate bone mineral density (BMD), but can also be determined by other techniques such as ultrasonic densitometry. Metastatic bone disease can be treated by some combination of surgery, chemotherapy, radiotherapy and radionuclide therapy.

**Related Articles:** Bone densitometry, Bone–soft tissue interface, Tc-99m-labelled bone imaging agents

**Bone densitometry** (Diagnostic Radiology) Bone densitometry is the measurement of bone density or bone mass associated with both the thickness and structure of bones. A primary application is to diagnose, evaluate and monitor osteoporosis, a common disease that increases the risk of bone fracture.

Bone density can be measured with several techniques including quantitative ultrasound (QUS), quantitative computed tomography (QCT) and single-energy x-ray absorptiometry (SEXA or SXA). However, the most common method for clinical bone densitometry is dual-energy x-ray absorptiometry (DEXA or DXA) – Figure B.67.

DEXA scans a selected region of the patient’s body with a small beam of x-rays (pencil shaped or cone shaped). The beam applies alternatively two different x-ray photon energy spectra (by using two different kVp). Because of the different x-ray attenuation characteristics for the two spectra in soft tissue and bone, the scan data can be processed to subtract out soft-tissue attenuation, leaving an image of the bones. Data from the bone images are used to calculate bone density values (Figure B.68).

**FIGURE B.67** Typical DEXA system (Hologic QDR4500) in a position for densitometry of the lumbar spine. (Image courtesy of Hologic, Bedford, MA.)

**FIGURE B.68** Part of a typical software interface used to calculate bone densitometry (sample from measurements and calculations in the region of the lumbar spine).
Bone–soft tissue interface

(Radiotherapy) There is an enhanced dose pattern at the interface of bone and soft tissue for photon beams due to

1. Backscattered photons
2. Backscattering of secondary electrons set in motion in the soft-tissue medium upstream of the bone
3. Backscattering of secondary electrons set in motion within the bone

It has been shown (Das and Khan, 1989) that the main component of this enhanced dose is due to the secondary electron backscattering at the interface. All dose enhancement is upstream of the bone. For a number of photon energies studied, from cobalt 60 to 24 MV, the range of the backscattered electrons upstream (beam entry side) is limited to a few millimetres and the magnitude of the dose enhancement effect (approximately 8%) is similar. On the transmission side of the bone, the forward scattered electrons from bone make the situation more complicated and energy dependent. Khan (2003) has shown that for energies less than 10 MV, there is an initial dose reduction at the interface relative to dose in a homogeneous soft-tissue volume. The dose then builds up to a value slightly higher than that for the homogeneous case. For higher energies, there is an enhanced dose at the interface due to electron fluence increase in bone resulting from pair production.

See Figure B.69 for an illustration of the dose pattern.


Boost (brachytherapy)

(Radiotherapy, Brachytherapy) Brachytherapy is often given as a boost together with external beam radiotherapy. Two examples of radiotherapy regimes with brachytherapy boosts are given.

---

EXAMPLE 1: LOCALISED HIGH-RISK PROSTATE CANCER

External beam radiotherapy is used to give 50 Gy in 25 fractions to the prostate with a ‘larger’ margin (conformal beam technique or IMRT technique). HDR interstitial brachytherapy is used to give a boost to the prostate with narrow margins, consisting of 2 fractions of 10 Gy each. The dose to organs at risk must be kept below accepted biological effect limits; urethra, rectum, etc.

EXAMPLE 2: CANCER OF THE CERVIX, RADICAL RADIOThERAPy

External beam radiotherapy with concomitant chemotherapy (chemoradiation) is used to give 46.8 Gy in 26 fractions to a larger target volume – pelvic lymph nodes including ‘uterus and tumour’, with an external beam boost of 3.6 Gy in 2 fractions to a smaller target volume – ‘uterus and tumour with margins’; in total, 50.4 Gy in 28 fractions is given to this volume. HDR intracavitary brachytherapy (ring applicator) is used to give a boost to the tumour with appropriate margins (see the GEC-ESTRO ref for a discussion of the brachytherapy target volume), consisting of 5 fractions of 5 Gy each. The dose to organs at risk, rectum, bladder, sigmoid colon, must be considered.

Abbreviations: HDR = High dose rate and IMRT = Intensity-modulated radiotherapy.

Related Articles: Intracavitary brachytherapy, Temporary implant, Interstitial brachytherapy


Boost dose

(Radiotherapy) Boost dose is defined as a supplemental dose of radiation delivered to the tumour bed following a conservative surgery in the treatment of the breast. The boost dose can be delivered either with an external radiation source and interstitial radioactive implant. In case of boost with an external irradiation, a smaller treatment field is used.

Born approximation

(Ultrasound) To evaluate the pressure scattered from a region of inhomogeneities, an integral equation needs to be solved. In this expression, the total pressure (i.e. both the incident and the scattered pressure) is needed, but this is unknown, since one is trying to calculate the scattered pressure. A common approximation is then to use the Born approximation, which is valid if the scattered pressure is much smaller than the incident pressure. Another way to put this is that the incident pressure is virtually unchanged as it passes through the volume in question, i.e. $p_{in} = p_i$. The scattered pressure will be small if the compressibility and density variations of the scattering objects are small compared to the surrounding medium. A consequence of this is also that multiple scattering (scattering of the scattered wave) can be ignored. By employing the Born approximation...
approximation, the scattered pressure can be evaluated directly, and not by using, for instance, successive approximation.

**Boron**  
*(General)*

<table>
<thead>
<tr>
<th>Symbol</th>
<th>B</th>
</tr>
</thead>
<tbody>
<tr>
<td>Element category</td>
<td>Metalloid (semimetal)</td>
</tr>
<tr>
<td>Mass number A</td>
<td>10, 11 (stable isotopes)</td>
</tr>
<tr>
<td>Atomic number Z</td>
<td>5</td>
</tr>
<tr>
<td>Atomic weight</td>
<td>10.811 g/mol</td>
</tr>
<tr>
<td>Electronic configuration</td>
<td>1s² 2s² 2p⁰</td>
</tr>
<tr>
<td>Melting point</td>
<td>2349 K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>4200 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>2.34 g/cm³</td>
</tr>
</tbody>
</table>

**History:** Boron was discovered jointly by Sir Humphry Davy, Joseph Louis Gay-Lussac and Louis Jacques Thénard in 1808.

**Isotopes of Boron:** Boron is found naturally only in compound form, notably as sodium borate (borax) and hydrated sodium calcium borate hydroxide (ulexite). Two stable isotopes exist: ¹⁰B and ¹¹B (19.9% and 80.1% relative abundance, respectively).

**Medical Applications:** Antiseptic – Diluted boric acid is sometimes used as a topical application for minor cuts, burns and wounds due to its anti-bacterial and antiseptic action.

Boron Neutron Capture Therapy (BNCT) – An application currently under investigation for the treatment of aggressive brain tumours. The therapy is delivered in two parts and utilises the high affinity of ¹⁰B for capturing thermal neutrons. An intravenous injection is administered containing a ¹⁰B-labelled agent, which is selectively taken up at the tumour site. The region is then irradiated with a beam of low-energy neutrons, which lose energy in tissue to become thermal neutrons and are absorbed by ¹⁰B in the tumour to form lithium ions and alpha particles:

\[ ^{10}\text{B} + ^{1}\text{n} \rightarrow ^{11}\text{B}^* \rightarrow ^{7}\text{Li} + ^{4}\text{He} \]

Both ⁷Li and ⁴He cause ionisations in tissue with a path length of 5–12 μm, approximately a single-cell diameter. This theoretically provides a significant dose to the tumour, with substantial reduction in collateral damage to healthy tissue compared with conventional radiotherapy techniques. However, the feasibility of this therapy has not currently been proven and is the subject of ongoing trials.

Neutron Shielding: Boron has a high thermal neutron absorption cross section, favouring its use as a shielding material in high-energy radiotherapy linear accelerator units. Boron is added to a liquid, usually paraffin, which is incorporated into the door of the linear accelerator maze to prevent the escape of potentially harmful neutrons.

**Related Article:** Neutron capture therapy

**Boron neutron capture**  
*(Radiotherapy)* Neutron capture is a nuclear reaction in which a neutron collides with an atomic nucleus and they merge to form a heavier nucleus. The incident neutron is swallowed up, giving rise to more or less stable nuclear excited states of the resulting isotope that may lead to an immediate de-excitation by the emission of a gamma photon as well as a delayed emission process when one or more nuclear particle may be emitted. The usual notation for the reaction is \((n,p), (n,n), (n,2n), (n,α)\) or \((n,\text{fission})\). The heavy charged particles produced in the nuclear reaction, i.e. proton and alpha particle, have a higher LET than gamma rays. The energy of the heavy particle is therefore distributed throughout a limited volume of approximately a sphere of 14 micron diameter, i.e. slightly larger than the diameter of a red blood cell. Boron neutron capture therapy (BNCT) is a modality of radiotherapeutic treatment of brain tumours that utilises a neutron beam that interacts with boron-10 (¹⁰B) injected to a patient. ¹⁰B, which is a stable isotope, absorbs a thermal neutron, producing the nuclear reaction \(^{10}\text{B}(n, α) ^{7}\text{Li}\). Alpha particles and lithium ions have a combined path length of approximately 12 μm and therefore deposit most of their energy within the cell containing the original ¹⁰B atom. If a high concentration of ¹⁰B is obtained in tumour cells, a high dose could be delivered to tumour cells, leaving normal cells unaffected. The interval between administration of the capture agent and neutron irradiation could be optimised to have the highest differential ¹⁰B concentrations between normal tissues and the tumour. Alpha particles and lithium ions will produce a significant radiobiological effect because of their high linear energy transfer (LET) as the particles experience closely spaced events. Alpha particles have another biological advantage as they do not require oxygen to enhance their biological effectiveness. This makes them very effective in cases where the tumour has limited oxygen supply. The neutrons are created either in a nuclear reactor or in particle accelerators, making a proton beam collide into targets made of lithium or beryllium. The neutrons pass through a moderator, which makes the neutron energy spectrum suitable for BNCT treatment, and a collimation device, which shapes the beam before entering the patient. Passing through the patient tissue, the neutrons are slowed by elastic scattering and become low-energy thermal neutrons. The thermal neutrons undergo reaction with the boron-10 nuclei, forming an excited compound nucleus boron-11 (¹¹B), which then promptly disintegrates to an alpha particle and ⁷Li. The investigation of carriers more efficacious than boron has recently addressed the use of ¹⁵⁷Gd. ¹⁵⁷Gd has a large thermal neutron cross section of 255,000 barn, which is 65 times that of ¹⁰B and releases Auger electrons, internal conversion electrons, gamma and x-rays by a single thermal neutron capture reaction sharing among them the total kinetic energy of 7.7 MeV, which is more than two times that of ¹⁰B(n, α)⁷Li reaction. In the BNCT, the heavy charged particles release 3.3 MeV only within their total trajectory. Such limited energy transfer within short trajectory in tissue can save serious radiation injury to the normal brain surrounding the tumour. However, dose distribution in the tumour is sharply dependent on the microdistribution of ¹⁰B in tumour that is heterogeneous and has non-uniform dose distribution, aside from a variety of proliferation conditions of the tumour cells. Dose distribution of Gd-based NCT is more uniform than that of boron-based NCT and might be suitable for pathological heterogeneity of malignant tumours.

**Related Article:** Linear energy transfer (LET)

**Boundary layer**  
*(Ultrason*) Boundary layer is a term used in fluid dynamics. When fluid flows over a surface, the fluid close to the surface moves at a velocity slower than that in the free stream. The thickness of the boundary layer is usually described as the distance from the surface to a point where velocity is 99% of free stream velocity (Figure B.70).

In Figure B.70, the shear stress at the wall causes the fluid at the wall to be stationary. This in turn creates a drag on adjacent fluid, slowing it. Further from the wall, the fluid flows at free stream velocity. The region of slower moving fluid is the boundary layer.

Boundary layers can be laminar or, at higher Reynolds numbers, turbulent. In turbulent boundary layers, flow is characterised by unsteady flow with vortices. In tubes, laminar flows are present at Reynolds numbers <2000 and turbulent >4000. Between these values, transition occurs.
The presence and extent of boundary layers in arteries and veins lead to complex velocity distributions, which are evident in the Doppler sonograms. One practical consideration in Doppler ultrasound quality assurance is that the flow profiles in a tube change as flow develops along it. This is shown diagrammatically in Figure B.71 where steady flow in a constant diameter straight tube only develops a parabolic flow profile when the boundary layer extends across the whole diameter. This is important if flow phantoms are to be used to produce a known parabolic flow profile where peak velocity $= 2 \times$ mean velocity. The entry length for flow to become established is approximately $120 \times$ diameter for laminar flow.

As flow moves down the pipe, the boundary layer (grey) increases until it extends across the pipe. At this point, flow is said to be fully developed and for constant laminar flow, has a parabolic flow profile.

Related Articles: Reynolds number, Laminar flow, Turbulent flow

**Bow-tie filter**

(Diagnostic Radiology) The bow-tie filter, also referred to as ‘beam shaping’ filter, or sometimes ‘wedge’, is a physical filter used on CT scanners to modify the x-ray beam profile in the scan (x–y) plane. It is usually employed in addition to the flat filter used to remove low-energy x-rays from the beam (Figure B.72).

The bow-tie filter compensates for the shape of the patient cross section by having a reduced thickness on the central axis of the beam where patient attenuation is highest. It provides additional beam hardening where the path of the beam through the patient is short, and so results in a more uniform beam energy at the detectors. The uniformity of dose and image quality across the patient cross section is also improved.

Some models of CT scanners will have two or more different bow-tie filters, designed for head and body applications or for patients of different sizes. More recently, some CT scanner models incorporate a special bow-tie filter for cardiac CT scans. These are shaped so as to reduce the dose, and thereby the image quality, in the area outside the central region of the beam as this is not of primary interest when imaging the heart.

**Related Article:** Beam hardening

**Boxcar function**

(Magnetic Resonance) Boxcar function or boxcar design is also known as block design. For further details, see Block design (MRI).

**Brachytherapy**

(Radiotherapy, Brachytherapy) Depending on the distance between the radiation source and the target volume, i.e. the tissues to be treated, radiotherapy is divided into two categories: teletherapy and brachytherapy:

- In teletherapy, the source is far from the target (Greek word ‘tele’, meaning distant, far away).
- In brachytherapy, the source is placed close to or inside the target (Greek word ‘brachys’, meaning short).

Brachytherapy conventionally uses sealed radioactive sources.

**Vocabulary in Brachytherapy**: There are many different ways of characterising brachytherapy (also called curietherapy), using words both from Latin and from Greek:

- In intracavitary brachytherapy, applicators/sources are placed in existing cavities (Latin words intra, meaning within, inside, and cavitus, meaning cavity, hole).
- In interstitial brachytherapy, applicators/sources are placed within the tumour using needles or catheters, ‘you make the cavities yourself’ (Latin words inter, meaning between, and interstitium, meaning gap, ‘a space between’).
- In endobronchial brachytherapy, applicators/sources are placed in the bronchus, a type of intracavitary technique (Greek word endon, meaning within).
- In endoluminal brachytherapy, the applicators/sources are placed within (Greek, ‘endon’) the lumen (Latin, meaning passage within a tubular organ); also a type of intracavitary technique. Using Latin (intra) instead of Greek (endo)
Brachytherapy sources can be divided into several different types as follows:

- Brachytherapy conventionally uses radioactive sources; thus, there is always a hazard of radiation exposure.
- Today, remotely controlled afterloading equipment is used for high-dose-rate (and pulsed-dose-rate) brachytherapy, and also for low-dose-rate brachytherapy, in principle eliminating the radiation hazard for the staff involved.
- Brachytherapy gives a high dose to a small target volume, with a rapid fall off outside the target, an advantage of brachytherapy. The steep dose gradient could on the other hand also lead to target under dosage at the periphery.
- Brachytherapy cannot be used to treat very large target volumes. Further, brachytherapy can only be used where the target volume is accessible for application/insertion of the appropriate applicators.
- The dose in the target volume is inherently very inhomogeneous in brachytherapy, with regions with doses larger than 50% and more of the peripheral dose. Thus, large parts of the target receive doses markedly higher than the prescribed dose. This is in contrast to the case in external beam therapy, where the dose to the target is almost constant, equal to the prescribed dose, with a variation about ±5%. For the same specified dose, brachytherapy gives large parts of the target volume a much higher dose than external beam therapy.
- Another noticeable difference between external beam radiotherapy and brachytherapy is the impact of patient motion, both external motions of the patient and motion of internal organs. This is a big problem in external beam therapy, and elaborate techniques are developed to handle different types of patient motions. In brachytherapy, the problem of these movements is minimised when the applicators/sources are placed directly in the target volume! However, permanently implanted sources can migrate within the target volume.
- In modern high-dose-rate and pulsed-dose-rate brachytherapy, it is possible to manipulate the dose distribution to match the target volume. ('Brachytherapy is the ultimate form of conformal radiotherapy!')
- Brachytherapy uses image-guided techniques today, Image-guided brachytherapy – IGBT, using ultrasound, CT, MR, fluoroscopy, etc.
- Brachytherapy is a radio-surgical procedure with special requirements for the radiation physics part of the treatment.
- Brachytherapy requires proper education and training for the whole brachytherapy team.

Different ways to characterise brachytherapy: Brachytherapy can be divided into several different types as follows:

1. According to the placement of the sources
   a. Intracavitary techniques
   b. Interstitial techniques
   c. Surface applications
2. According to dose rate
   a. Low dose rate – LDR
   b. Medium dose rate – MDR
   c. High dose rate – HDR
   d. Pulsed dose rate – PDR
3. According to duration of treatment
   a. Temporary implants
   b. Permanent implants
4. According to the handling of the sources
   a. Manual handling
   b. Manual afterloading
   c. Remotely controlled afterloading
5. According to radiation quality
   a. Sources emitting mainly photons
   b. Sources emitting beta radiation
   c. Sources emitting neutron radiation

These types are presented in some more detail under related articles.

**Related Articles:** Intracavitary brachytherapy, Intraluminary brachytherapy, Intravascular brachytherapy, Interstitial brachytherapy, Dose rates in brachytherapy, Temporary implant, Permanent implant, Source loading in brachytherapy, Brachytherapy sources


**Brachytherapy sources**
*(Radiotherapy, Brachytherapy)*

**Introduction:** When brachytherapy was introduced, the naturally occurring radioactive isotopes Radium-226 and Radon-222 were used as photon-emitting brachytherapy sources. Later, particle accelerators and nuclear reactors made it possible to produce new radionuclides with a wide range of physical properties. Today, there are a number of radionuclides used in brachytherapy, emitting photon, beta and neutron radiations.

The historical radium and radon sources are in principle not used any longer, and should not be used, primarily because of safety concerns. However, there is a large amount of clinical experience gained using radium sources and specific treatment techniques, which must not be forgotten.

Some photon-emitting nuclides are denoted ‘radium substitutes’, for example caesium-137, iridium-192 and cobalt-60, as the dose rate distributions for these sources and for radium sources are very similar up to about 5 cm distance from the source. Originally, source strengths were specified in terms of ‘mg Ra eq’ (mg radium equivalent).

**Source Characteristics:** Brachytherapy sources are in general encapsulated; the radioactive material is contained in a capsule often made of stainless steel or titanium (non-toxic materials). The capsule further serves as an absorber for ‘unwanted decay products’, such as alpha particles and for photon sources also beta radiation. Modern sources are small, and generally cylindrical in shape.

Photons-emitting sources are by far the most commonly used sources for all types of brachytherapy. Beta-emitting sources are used for the treatment of superficial tumours/lesions, for example in ophthalmic applicators (and in intracoronary brachytherapy).
Properties of some brachytherapy sources are given as follows; the energy values – nominal energy values, principal or mean values – are given for a typical encapsulated source, half value layers are given for lead and the mass values in the last column are theoretical calculated maximum values.

<table>
<thead>
<tr>
<th>Radionuclide</th>
<th>Type</th>
<th>Half-Life</th>
<th>Energy (MeV)</th>
<th>HVL (mm Pb)</th>
<th>Mass for 100 MBq (µg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ra-226, historical</td>
<td>Photon</td>
<td>1600 years</td>
<td>0.83</td>
<td>16</td>
<td>2732</td>
</tr>
<tr>
<td>Ra-226, historical</td>
<td>Photon</td>
<td>3.82 days</td>
<td>0.83</td>
<td>16</td>
<td>0.02</td>
</tr>
<tr>
<td>Co-60</td>
<td>Photon</td>
<td>5.27 years</td>
<td>1.25</td>
<td>11</td>
<td>2.39</td>
</tr>
<tr>
<td>Cs-137</td>
<td>Photon</td>
<td>30.07 years</td>
<td>0.662</td>
<td>6.5</td>
<td>31.1</td>
</tr>
<tr>
<td>Ir-192</td>
<td>Photon</td>
<td>73.83 days</td>
<td>0.380</td>
<td>3.0</td>
<td>0.29</td>
</tr>
<tr>
<td>I-125</td>
<td>Photon</td>
<td>59.4 days</td>
<td>0.028</td>
<td>0.025</td>
<td>0.15</td>
</tr>
<tr>
<td>Pd-103</td>
<td>Photon</td>
<td>17.0 days</td>
<td>0.021</td>
<td>0.013</td>
<td>0.04</td>
</tr>
<tr>
<td>Sr-90/90Y</td>
<td>Beta</td>
<td>28.8 years</td>
<td>0.55−2.28</td>
<td>0.14</td>
<td>19.6</td>
</tr>
<tr>
<td>(Cf-252)</td>
<td>Neutron</td>
<td>2.65 years</td>
<td>2.15</td>
<td>—</td>
<td>—</td>
</tr>
</tbody>
</table>

Some characteristics to consider in the choice of a photon-emitting brachytherapy source for a clinical application are

1. Energy
   a. ‘Low’; minimises radiation protection requirements
   b. ‘Reasonably high’ (radium substitute sources)
2. Half-life
   a. Longer; temporary implant sources – long working life
   b. Shorter; sources for permanent implants
3. Specific source strength
   a. High; small source dimensions and small applicator dimensions
4. Toxicity (problems with radium)
5. Decay products
   a. No gaseous products (problem with radium–radon)
   b. No electrons that are not absorbed in the encapsulation

Brachytherapy using radionuclide sources like Iridium-192 for temporary implants and Iodine-125 for permanent implants is a well-established clinical procedure.

**Developments:** The brachytherapy sources mentioned in the earlier table are not the only sources that have been used for brachytherapy; other sources are, for example Tantalum-182, Ruthenium-106, Americium-241, Samarium-145 and Ytterbium-169.

There are a large number of radionuclides, both naturally occurring and artificially produced, that could be used for different types of brachytherapy, and developments of new sources and applications are continuously ongoing.

Other interesting developments are the brachytherapy sources that do not use radionuclides for the production of photons, consisting of miniature x-ray sources.

**Related Articles:** Brachytherapy, Radium, Radium substitute isotope, Iodine-125, Iridium-192, Intravascular irradiation, Equivalent mass of radium

**Bragg peak**

(Radiotherapy) Heavy charged particles such as protons or heavier particles interacting with a medium show a characteristic-shaped depth-dose distribution, if their nuclear interactions in the medium are negligible. If some of the particles will be subject to inelastic nuclear reaction before they have come to rest, lighter fragments with longer ranges than that of the primary ion are produced, depositing the dose observed beyond the Bragg peak. As the charged particle proceeds through a medium, its rate of energy loss or ionisation per unit path length increases with decreasing particle velocity until it reaches a maximum in correspondence of the Bragg peak near the end of its path. The shape of the depth-dose curve is determined primarily by the type of ion and its initial energy spectrum. The shape of the distribution is due to the fact that the heavy charged particle spends the first half of its initial kinetic energy along a pathlength x and the remaining half of the energy will be spend in a distance roughly equal to x/3, increasing in this way the rate of energy deposition at the end of its track. Using relativistic quantum mechanics, the stopping power of a uniform medium for a heavy charged particle is given by

\[-\frac{dE}{dx} = \frac{4\pi k_n^2 z^2 e^4 n}{m c^2 \beta^2} \ln \frac{2mc^2\beta^2}{I(1-\beta^2)} - \beta^2\]

where

- \(k_n\) is the 8.99 × 10⁹ N m²/C²
- \(z\) is the atomic number of the heavy particle
- \(e\) is the magnitude of the electron charge
- \(n\) is the number of electrons per unit volume in the medium
- \(m\) is the electron rest mass
- \(c\) is the speed of light in vacuum
- \(\beta = v/c\) is the speed of the particle relative to \(c\)
- \(I\) is the mean excitation energy of the medium

At high energies when \(\beta \approx 1\), the logarithmic term makes the stopping power increase. At low energies, the factor in front of the bracket increases as \(\beta \approx 0\) and the logarithm term then decreases, causing the so-called Bragg peak. The linear rate of energy loss of the charged particle is a maximum there. The shape of depth–dose distribution of the heavy charged particle makes possible the irradiation of a strictly localised region within a biological object in correspondence of the Bragg peak depth. In **Figure B.73**, the Bragg peak of a 62 MeV proton beam in water is shown.

The exact location of the Bragg peak can be measured with use of the Bragg peak chamber. The diameter of the chamber is large enough to measure the proton beam diameter.

**Related Articles:** Stopping power, Deposition of dose

**FIGURE B.73** The Bragg peak (62 MeV protons).
Bragg–Gray cavity theory

(Radiotherapy) The Bragg–Gray theory considers a medium uniformly irradiated by photons and whose dimensions are adequate to establish a condition of electronic equilibrium in which a small gas-filled cavity is inserted. The theory relates \( D_m \), the absorbed dose in the medium, to \( J \), the charge per unit of mass resulting from ionisation produced by electrons in the gas. If the cavity gas is air, \( D_m \) is given by

\[
D_m = J_{air} \left( \frac{W}{e} \right) \left( \frac{S}{\rho} \right)_{m,air}
\]

where

- \( J_{air} \) is the ionisation charge per unit mass of air in the cavity
- \( (S/\rho)_{m,air} \) is the ratio of the mean unrestricted collision mass stopping power of material \( m \) to that of air
- \( W/e \) is the quotient of the average energy expended to produce an ion pair by the electronic charge

The theory requires that

- Charged particle equilibrium exists in the absence of the cavity
- The cavity does not disturb the charged particle fluence or its distribution in energy and direction
- The mass stopping power ratio does not vary with energy
- Secondary charged particles lose energy by a process of continuous slowing down

The theoretical approach to Bragg–Gray dosimetry requires the cavity and therefore the radiation detector so small that, when inserted into a medium, it does not disturb the fluence of charged particles existing in the medium. This means that the ideal Bragg–Gray cavity is one of infinitesimal dimensions and therefore the detector must be a point detector to fulfil the hypothesis of the theory. In practice, such detectors do not exist but many real detectors may, in a first approximation, be treated as Bragg–Gray detectors to a high degree of accuracy. However, perturbation corrections are needed to account for the deviation of the signal from a practical detector from that of an ideal.

**Related Article:** Stopping power

Braking radiation

(Radiation Protection) Braking radiation is more usually called Bremsstrahlung. For more information, see eponymous article.

**Related Articles:** Bremsstrahlung, Characteristic x-rays

Breakdown voltage

(General) Breakdown voltage of an insulator is the minimum voltage that causes a portion of an insulator to become electrically conductive.

**Breakdown voltage** of a diode is the minimum reverse voltage to make the diode conduct in reverse, as described later. Some devices (such as TRIACs) also have a forward breakdown voltage.

**Figure B.74** shows the dependence of the current flow on the voltage for a diode \((p–n \text{ junction})\). This \( p–n \) junction characteristic is nonlinear.

The current flowing across the \( p–n \) junction depends on the direction of the potential difference (bias). The current increases with the voltage increasing in a forward direction \((\text{OF})\). If the normal reverse bias is applied to the diode, the low reverse current \((\text{OR})\) flows due to minority carriers. The diode can be used as rectifier.

The further increase of the reverse voltage makes the diode conduct in reverse \((RB)\). The diode stops to act as a rectifier. The minimum reverse voltage causing that the reverse current increases strongly is called breakdown voltage. At this voltage, the minority carriers are accelerated to get enough energy to ionise molecules. As a result of this ionisation, the number of charge carriers (electrons) increases and the reverse current grows. It is the so-called solid-state multiplication.

**Abbreviation:** TRIAC = Triode for alternating current.

**Related Article:** Semiconductor detector


Breast

(General)

<table>
<thead>
<tr>
<th>Density</th>
</tr>
</thead>
<tbody>
<tr>
<td>( \sim 1000 \text{ kg/m}^3 )</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>CT number</th>
</tr>
</thead>
<tbody>
<tr>
<td>(-100 \text{ to } 0 \text{ HU})</td>
</tr>
</tbody>
</table>

The breast is the upper ventral region of the torso containing modified sweat (mammary) glands in females, which lactate (secrete milk) for breast-feeding. Estrogens in females promote breast development. Each breast has a nipple surrounded by the areola, which have sebaceous glands, and the mammary glands drain to the nipple via a lactiferous duct network. The main structure of the breast consists of connective tissue, adipose tissue and Cooper’s ligaments, and the pectoralis major muscle lies underneath the breast. The lymphatic drainage of breasts predominantly travels through the ipsilateral axillary lymph nodes, which is pertinent to oncology since breast cancer is common and metastatic spread occurs via the lymphatic system.

Breast tissue is assessed in medicine usually for oncological purposes, predominantly by mammography, as used in the breast screening programme in the United Kingdom. The breast may also be assessed by ultrasound, MRI and nuclear medicine sentinel node scans to locate lymph nodes for biopsy. Breast cancer is treated by some combination of surgery, chemotherapy and radiotherapy.

**Related Articles:** Digital mammography, Mammography, Radiotherapy, Ultrasound
Breast coil

(Magnetic Resonance) A breast coil is used in MRI as the RF coil for breast imaging. During imaging, the patient lies prone on the MRI table. Recesses in the breast coil housing accept each breast and adjustable paddles may be used to hold the breasts in position (Figures B.75 and B.76).

Bremsstrahlung

(Radiation Protection) Bremsstrahlung is a German word that means ‘braking radiation’.

When the electrons are incident upon the anode target, they are rapidly decelerated by interactions with the electric fields within the atoms of the target. There are two mechanisms by which the electrons are decelerated (Figure B.77).

Firstly, incident electrons may interact with the electric field of the nuclei. The path of the incident electron is changed by this interaction, implying that the electron is decelerated – i.e. braked, giving off energy in the form of x-ray photons. Because there are any number of different paths by which the electrons may traverse through the atoms of the material, the x-rays emitted will have a spectrum of energies. This continuous spectrum of x-rays is therefore known as Bremsstrahlung, or braking radiation.

Secondly, they may interact with the electric field of an inner bound atomic electron, knocking that electron out of its atomic orbit, and losing some or all of its kinetic energy in the process. If not all its energy is lost, the incident electron is deflected by the collision. The vacancy in the inner orbit is filled by an electron dropping down from an outer orbit. The electron must emit energy in the form of an x-ray photon. The energy of the photon is equal to the energy gap between the orbits, and as such is characteristic of the atom of the material – hence these x-rays are known as characteristic x-rays.

Related Articles: Braking radiation, Characteristic x-rays

Bremsstrahlung contamination

(Radiotherapy) The electron beams from a linac always have some x-ray photons present, and these are known as Bremsstrahlung contamination. They are produced in the head of the linac and result in a Bremsstrahlung tail being present in electron percentage depth-dose plots. This contamination is typically of the order of 3% (Figure B.78). Knowledge of the Bremsstrahlung contamination of an electron beam is also of importance for a specialist treatment known as total skin electron therapy.

Abbreviation: PDD = Percentage depth dose.

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Related Articles: Braking radiation, Characteristic x-rays

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Abbreviation: PDD = Percentage depth dose.
**Related Article:** Electron ranges


**Brick**  
*(Radiation Protection)* See Radiation shielding

**Bridge circuit**  
*(General)* A bridge circuit is a term commonly used to describe a specific architecture of four components.

The bridge rectifier is a commonly available unit formed from four diodes, and used to convert AC signals or power to DC (Figure B.79).

The impedance bridge is designed to provide an output signal, which is related to the RATIO of impedances of the conductors while being relatively insensitive to changes in the supply voltage or ambient temperature (Figure B.80).

When used with a sensitive current meter across the output arms, it becomes a traditional method for determining the unknown impedance of a component: one half is made of two identical impedances $Z_1, Z_2$, the third is calibrated variable impedance $Z_3$ and the fourth is the unknown $Z_4$. The variable impedance $Z_3$ is merely adjusted to nullify any current flowing in the current metre, at which point the variable impedance MUST equal the impedance of the unknown component. It has the two advantages of being a highly sensitive circuit whilst being insensitive to the amplitude of the signal applied to power the circuit or any parameter that affects the components similarly (e.g. temperature change).

**Related Articles:** Diode, Rectifier, Four rectifier circuit


**Brightness**  
*(Diagnostic Radiology)* Brightness is the characteristic of a light source or reflecting surface describing the visual response or perception related to the amount of light emitted.

It is related to the measurable physical quantity luminance, which is the density of luminous intensity emitted in a specific direction. The SI unit for luminance is the candela per square metre (cd/m²).

In diagnostic radiology, the brightness (or luminance) of images (film on illuminators and digital displays) has an effect on visual perception and must be adjusted to an appropriate level.

When the perceived power of light is considered, then luminous flux is used – measured in Lumen (lm) – an SI unit.

$$1	ext{ lm} = 1 \text{ lx} \cdot \text{m}^2 = 1 \text{ cd} \cdot \text{sr}$$

Here lux (lx) is the SI unit of illuminance and luminous emittance.

$$1 \text{ lx} = 1 \text{ lm}/\text{m}^2$$

Lux is used for light incident on a surface, while Lumen is used for light emitted from a surface.  

**Hyperlink:** [http://en.wikipedia.org/wiki/Lumen_(unit)]

**Brightness control**  
*(Diagnostic Radiology)* See Automatic brightness control (ABC)

**Brightness induction effect**  
*(Diagnostic Radiology)* Brightness induction is the apparent difference in the brightness or intensity of an object when the background is changed from light to dark as illustrated in Figure B.81.

**Brightness stabilisation**  
*(Diagnostic Radiology)* See Automatic brightness control (ABC)

**Broad-beam geometry**  
*(Nuclear Medicine)* Broad-beam geometry refers to situations in which the radiation beam profile is larger than a few cm². The opposite situation is referred to as narrow-beam geometry. Consider two different beam geometries striking a perpendicular surface. The dose along the central line in the two cases will differ since in the broad-beam situation, photons outside the central line can scatter inwards and contribute to the dose. The dose along a central line is therefore higher in the broad-beam geometry compared to a narrow beam with equal intensity per unit area.

**Related Articles:** Narrow beam, Absorber, Broad-beam geometry

**B-scanner**  
*(Ultrasound)* Basic diagnostic ultrasound imaging (greyscale images) is performed with a B-scanner. Other common words for ultrasound imaging instrumentation are linear scanner, ultrasound

**Related Articles:** Narrow beam, Absorber, Broad-beam geometry

![Figure B.79](image1.png)  
**Figure B.79** The bridge rectifier.

![Figure B.80](image2.png)  
**Figure B.80** The impedance bridge.
Bucky diaphragm


Bucky diaphragm

Bucky diaphragm is the classic name for what is more commonly known as a grid used to selectively absorb scattered radiation in x-ray imaging. It is named after Dr. Gustav Bucky from Germany who developed the first grid or ‘diaphragm’ in 1913.

A major advancement was made by Dr. Hollis Potter in 1920, who developed a method for moving the grid during the exposure to blur out the undesirable images of the grid lines. This was known as the Potter–Bucky diaphragm.

The use of the names has been changed over the years. Now the ‘Bucky diaphragm’ is most commonly known as a grid and the ‘Potter–Bucky diaphragm’ is known as a ‘Bucky’ or a Bucky mechanism.

The application of a Bucky diaphragm or grid is illustrated in Figure B.83.

The grid is composed of parallel metal strips, such as lead, that attenuate the radiation. They are separated by spaces or gaps filled with a low-attenuating material, such as aluminium or fibre. In many grids, the strips (spaces) are aligned with or focused on a point in space, the grid focal point. The grid is positioned so that the primary x-ray beam is aligned with the strips (and interspaces) and can pass...
through or penetrate the grid with relatively little attenuation. This occurs when the grid focal point is located at the x-ray tube focal spot. Because the patient’s body, which is the source of the scattered radiation, is much closer than the x-ray tube to the grid, the scattered radiation is not aligned with the interspaces and much of it is attenuated.

The desirable characteristic of a grid is to attenuate as much of the scattered radiation as possible and let the primary beam pass through.

An important performance variable of a grid is the grid ratio (usually with values from 5:1 to 16:1), i.e. ratio between height (thickness) of the strips and width of the interspace. A focused grid is also characterised by the focal distance (usually with values from 70 to 180 cm), but in practice it is usable in a focal range (e.g. grid with focal distance of 100 cm is usable in a range from 90 to 110 cm) without causing noticeable image non-uniformity.

**Related Articles:** Grid, Grid ratio, Focused grid

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**Bucky table**

_(Diagnostic Radiology)_ A Bucky table is a radiographic table with a built-in 'Bucky' or mechanism for moving the grid during the exposure.

**Related Articles:** Bucky diaphragm, Grid

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**Bucky wall stand**

_(Diagnostic Radiology)_ A Bucky wall stand is a stand for the cassette holder or for the flat panel detector with a moving grid mounted usually on the wall of a radiography room. It is most often used in chest radiography.

**Related Articles:** Bucky diaphragm, Grid

---

**Build-up**

_(Radiotherapy)_ At megavoltage energies, the scattered radiation is more in the forward direction and gives rise to less scattered radiation outside the edges of the beam. This is clear from a comparison of the isodose charts for kilovoltage and megavoltage beams. The Compton scattering process causes the recoil electrons to be ejected more in the forward direction and with increasing kinetic energy. The range of the recoil electron at megavoltage energies is on the order of a few millimetres, and an electron dose build-up to \( d_{\text{max}} \) manifests itself below the surface of the irradiated tissue, see Figure B.84.

**Related Articles:** Percentage depth dose, Build-up dose

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**Build-up cap**

_(Radiotherapy)_ If a dose chamber is to be used for exposure measurements under electron equilibrium conditions in a beam of energy \( \geq 0.5 \) MV, then an appropriate build-up cap must be used. Figures B.85 and B.86 show this.

**Related Articles:** Build-up, Ionisation chamber


**Build-up dose**

_(Radiotherapy)_ This may be explained in simple terms as follows. Each successive thin layer of tissue produces its quota of recoil electrons, which in turn deposit their kinetic energy through several successive layers of tissue beyond their point of origin. Although the kinetic energy released in each layer (the kerma) may be constant, the energy deposited in each layer (the absorbed dose) will be determined by the total number of electrons passing through the layer, and that number increases as each layer adds its quota of recoil

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**FIGURE B.84** Central axis dose build-up and fall-off for different kV and photon beam energies.

**FIGURE B.85** Farmer chamber and Perspex build-up cap.

**FIGURE B.86** Farmer chamber with build-up cap fitted.
electrons to the electron flux from the preceding layers. This number only continues to increase until the new electrons released only replace those that have come to the end of their range. The electron dose build-up thus reaches a maximum at a depth determined by the range of the electrons and therefore by the energy of the photon beam. In practice, the kerma is not quite constant but falls, as the primary radiation undergoes absorption and attenuation, with the result that the depth of the dose maximum is less than the maximum range of the secondary electrons; \( d_{\text{max}} \) is further reduced because the recoil electrons do not all travel in the direction of the primary radiation.

**Related Articles:** Build-up, Kerma


### Build-up plates

*(Radiotherapy)* In total body irradiation (TBI) treatments, better dose homogeneity throughout the patient is usually achieved using high-energy photon beams. However, the build-up effect of the percentage depth dose for these beams means that superficial or shallow structures close to the skin may not receive the prescribed dose. To get higher dose to these regions, build-up plates made from Perspex or similar material can be placed between the patient and the radiation source, close to the patient. These plates, which are usually 1–2 cm thick, provide a source of forward-scattered electrons when irradiated with high-energy photon beams. These electrons then increase the dose to shallow structures and improve dose uniformity in the patient.

**Related Articles:** Build-up, Build-up dose, Skin sparing


### Build-up region

*(Radiotherapy)* One of the benefits of using high-energy megavoltage beams instead of kilovoltage beams is that there is a skin sparing effect due to the fact that the dose at the skin surface is much lower than the dose to tissue at shallow depths. The region on the depth-dose curve where the dose increases from the skin surface to the depth of maximum dose is called the build-up region. The build-up region results from the range of secondary electrons produced in the patient/phantom by photon interactions – that is they deposit their kinetic energy beyond the position at which they are released. The amount of material contained within the build-up region as well as the significance of the build-up effect will be dictated by the energy and nature of the beam – see Table B.4 for a list of examples, Figure B.87 for an illustration of typical build-up region data for x-ray beams and Figure B.88 for electron beams.

While the absorbed dose build-up region is similar in appearance for both megavoltage electron and photon beams, the physical reason and size are different:

1. In the case of x-ray beams, the initial electron fluence is low, but as an increasing number of secondary electrons are produced, with depth the dose increases. There is then a point (depth of maximum dose, \( d_{\text{max}} \)) where the fluence of these electrons is matched (in equilibrium) with the exponential decrease in x-ray photons, and the dose delivered then starts to decrease with depth.
2. For electron beams, the deposition of energy (dose) begins immediately at the surface, and therefore the skin-sparing effect is reduced compared to x-ray photon beams. As the electrons penetrate further in the material, they are scattered at increasingly oblique angles, therefore increasing the fluence and thus dose delivered.

The significance of the build-up region depends on the energy of the beam since it is much easier to scatter a low-energy electron, and so a much more oblique pathway will result. Whereas higher energy electrons are less easily scattered and so tend to keep closer to a straight path, resulting in a much reduced build-up effect.

**Related Articles:** Build-up, Build-up dose, Skin sparing

### Build-up plates

*(Radiotherapy)*

**TABLE B.4**

<table>
<thead>
<tr>
<th>Treatment Beam</th>
<th>Surface Dose (%)</th>
<th>Build-Up Thickness (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>6MV x-ray</td>
<td>50</td>
<td>1.5</td>
</tr>
<tr>
<td>15MV x-ray</td>
<td>30</td>
<td>3.0</td>
</tr>
<tr>
<td>4MeV electrons</td>
<td>78</td>
<td>0.8</td>
</tr>
<tr>
<td>12MeV electrons</td>
<td>89</td>
<td>2.4</td>
</tr>
<tr>
<td>16MeV electrons</td>
<td>95</td>
<td>3.2</td>
</tr>
</tbody>
</table>

**Bulseye image** *(Nuclear Medicine)*

A polar map combines all areas of the myocardium into a 2D image and is standard in the most commonly used quantification systems. The image is created by wrapping each short axis image around the previous, starting from the apex. This type of image usually displays the raw counts, areas of abnormality and reversibility by combining and comparing the circumferential profile of each short axis slice into a colour-coded image and a gender-specific normal database. The points of each circumferential profile are assigned a colour based on normalised count values, and the colour profiles are shaped into concentric rings as shown in Figure B.89.

The slice-based polar map is either distance-weighted to adequately localise defects or volume-weighted to equally map areas of the myocardium to determine as accurately as possible the size of defect, though this defect size is distorted. Although this approach to perfusion quantification was a preferred option when nuclear medicine computers were slow, more realistic 4D gated SPECT displays are growing in popularity. Despite this, polar map displays are still frequently used.

**Related Articles:** Extent, Severity, Gated SPECT

### Bus

*(General)* The data bus in a computer system is the pathway between the central processing unit (CPU) and a peripheral device. Usually, contemporary buses have parallel data lines (most often 32 and 64 bits) between the CPU and any peripheral controller (card). The buses have also external ports for cables, connecting the computer with a specific device. Each data bus has its own speed of data transfer (MB/s), for example

- PCI – Peripheral component interconnect bus – 32 bits; 130MB/s
- AGP – Accelerated graphic port – 32 bits; at least 500MB/s
- USB – Universal serial bus – 480MB/s

**Hyperlink:** [www.pcmag.com/encyclopedia_term/](http://www.pcmag.com/encyclopedia_term/)

### b-Value

*(Magnetic Resonance)* The *b*-value is a factor describing the amount of weighting in diffusion-weighted MRI. In particular, it describes the strength of the diffusion encoding, and is the summarised influence of all gradients on the diffusion encoding. It describes the relationship

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**Hyperlink:** [www.pcmag.com/encyclopedia_term/](http://www.pcmag.com/encyclopedia_term/)
between the diffusion coefficient, $D$, and the signal attenuation. The higher the $b$-value is the stronger the diffusion encoding will be.

For the Stejskal–Tanner experiment (Stejskal and Tanner, 1965), the $b$-value is defined by

$$b = \frac{\gamma^2 G^2 \delta^2 (\Delta - \delta/3)}{\text{s/mm}^2}$$

where
- $\gamma$ in [Hz/T] is the gyromagnetic ratio
- $G$ in [T/m] is the amplitude of the diffusion encoding gradient
- $\delta$ in [ms] is the duration of the diffusion encoding gradient
- $\Delta$ in [ms] is the time interval between the onset of the two gradient pulses

The $b$-value is often determined only from the applied diffusion encoding gradients, but in order to accurately estimate the diffusion sensitivity, all gradients within a pulse sequence (imaging and slice selection gradient) should be included. To correctly determine the $b$-value of a pulse sequence, the time integral given later, including all gradient pulses, should be solved

$$b = \gamma^2 \int_0^{\tau_E} \left( \int_0^\tau G'(t) dt \right)^2 d\tau$$

where
- $G'$ is the amplitude of the gradient
- $\tau$ is the duration of the gradient (denoted $\delta$ for the diffusion encoding gradient)

See Figure B.90.
Bystander effects

Related Articles: ADC, Diffusion encoding, Diffusion time


Bystander effects

(Radiation Protection) Radiation-induced bystander effects are biological responses in non-irradiated cells, i.e. no energy deposition by the radiation in the cell. These cells are often adjacent to irradiated cells, hence the name bystander effects. Bystander effects are believed to be communicated via cell-to-cell junctions or by secretion or shedding of soluble factors. The bystander effect is proportional to the absorbed dose delivered to the targeted cells. The effect in a bystander cell can either be inhibitory (more radiosensitive) or stimulatory (cells tend to be less radiosensitive) (Figure B.91).

The first report of this phenomenon involved the exposure of a monolayer of cells to very low fluences of α-particles such that no more than 1% of cells were actually traversed by α-particles. The observation that changes were seen in 30%–50% of cells showed that the target for genetic damage by α-particles is much larger than the nucleus or indeed the cell itself.

There are also reports that chromosomal activity may be inherited by the clonal descendants of bystander cells.

The bystander effect is becoming better understood, and there is evidence that it can be attributed to inter-cell interactions and the release of factors into the media supporting the cell cultures.

As can be seen in the previous diagram, three cells are irradiated. The middle cell exhibits the classical direct damage leading to a mutation. The cell on the left only expresses the damage in the form of genomic instability observed several generations later. The cell on the right is irradiated, but passes on the damage to another cell by some form of chemical messenger. It is then an un-irradiated cell that expresses the damage. Various theories have been postulated why the irradiated cell might do this; the most obvious is as a survival mechanism.

However, there are two key implications of the bystander effect:

Firstly, the target volume for damage in tissue will be greater than the irradiated volume. Secondly, the response of the body to small doses of radiation may be higher than the linear no-threshold (LNT) model might suggest. This can be represented in the graph in Figure B.92:

The graph demonstrates that lower doses lead to a disproportionately higher rate of cancer induction than might be assumed using the classical LNT model for radiation protection.

Related Articles: Radiobiological models, Linear no-threshold model, Adaptive responses and hormesis


FIGURE B.90 A basic diffusion-sensitive pulse sequence, known as the Stejskal–Tanner pulse.

FIGURE B.91 Examples of bystander effects.

FIGURE B.92 Theoretical dose response assuming bystander effects.
**CAD (computer-aided diagnosis)**

*General*  See *Computer-aided diagnosis (CAD)*

**Cadmium tungstate**

*(Diagnostic Radiology)* Cadmium tungstate (CdWO$_4$) is a phosphor material used as a scintillator in some radiation detectors with photomultiplier tubes (especially in CT scanner detectors). Cadmium tungstate has conversion efficiency on the order of 40% (while CsI has ~45 and CaWO$_4$ has ~15). It emits light with peak wavelength of 520 nm.

**Related Articles:** CaWO$_4$, Caesium iodide


**Caesium**

*General*

<table>
<thead>
<tr>
<th>Property</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Symbol</td>
<td>Cs</td>
</tr>
<tr>
<td>Element category</td>
<td>Alkali metal</td>
</tr>
<tr>
<td>Mass number $A$</td>
<td>55</td>
</tr>
<tr>
<td>Atomic number $Z$</td>
<td>133</td>
</tr>
<tr>
<td>Atomic weight</td>
<td>10.811 g/mol</td>
</tr>
<tr>
<td>Electronic configuration</td>
<td>$1s^2\ 2s^2\ 2p^6\ 3s^2\ 3p^6\ 3d^{10}\ 4s^2\ 4p^6\ 5s^2\ 4d^{10}\ 5p^6\ 6s^1$</td>
</tr>
<tr>
<td>Melting point</td>
<td>301.59 K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>944 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>1.93 g/cm$^3$</td>
</tr>
</tbody>
</table>

**History:** Caesium was discovered by Robert Bunsen and Gustav Kirchhoff in 1860, through spectroscopic investigation of mineral water, in which it was seen as characteristic blue lines. Carl Setterberg first produced the metal through electrolysis of caesium mineral water, in which it was seen as characteristic blue lines. Carl G.

**Isotopes of Caesium:** Caesium is most commonly found in compound as a chloride or nitrate. It occurs as stable $^{133}$Cs with 100% natural abundance. There are 38 other known isotopes with atomic numbers between 112 and 151, all of which are unstable and decay through radioactive processes. The main isotope of interest in medicine is $^{137}$Cs, which decays via pure beta emission to $^{137m}$Ba with a half-life of 30.23 years. The $^{137m}$Ba created in this decay is often in an excited state and emits a gamma photon to reach its ground state with a half-life of 2.55 min, so that a sample of $^{137}$Cs emits both beta and gamma radiation.

**Medical Applications:** Nuclear Medicine Transmission Source and Marker – $^{137}$Cs is commonly used as a transmission source in nuclear medicine scanning, to provide information for the attenuation correction of SPECT and PET scans. The $^{137}$Cs source is positioned behind the object being scanned, and the gamma emissions are detected from the opposite side of the object to create an image whose intensity is proportional to the transmission of gamma rays through the tissue. If this is performed for each view acquired in emission, the transmission information can be incorporated into reconstruction calculations to correct for the attenuation of gamma photons with increasing depth in the tissue.

**Scintillation Detectors – Caesium fluoride was used as a scintillation detector material in early time-of-flight (TOF) PET imaging, as its high temporal resolution allowed improved localisation of radioactivity within the body. It has since been replaced with lutetium oxyorthosilicate (LSO) and lutetium–yttrium oxyorthosilicate (LYSO), both of which offer increased detection efficiency and spatial resolution as well as high temporal resolution.

**Caesium iodide doped with thallium (Cs(Tl)) is a common phosphor material used in digital radiography detectors, fluoroscopy image intensifiers and gamma cameras. The crystal produces flashes of visible light in response to incident gamma and x-ray photons to enable detection of the radiation.**

**Teletherapy – Radioactive $^{137}$CsCl was used in the past as a radiation source in caesium teletherapy units. The use of caesium for teletherapy was discontinued because, in comparison to cobalt-60, (Co-60) caesium-137 (Cs-137) has a relatively low specific activity (86.7 Ci/g) and relatively low energy (662 keV). Moreover, CsCl is a salt, highly dispersible in the atmosphere and soluble in water, making it very dangerous from the point of view of source security and disposal, in comparison to cobalt-60 which is a metal. Because of these drawbacks, Cs-137 is no longer used for radiotherapy; however, it is still used in blood irradiators, despite security concerns, because its long half-life of 30 years and relatively low gamma ray energy make it more practical than Co-60 for this purpose.

**Related Articles:** SPECT (single photon emission computed tomography), PET (positron emission tomography), Caesium fluoride, Caesium iodide, Caesium unit

**Caesium fluoride (CsF)**

*(Diagnostic Radiology)* Caesium fluoride (CsF) belongs to the chemical family of inorganic fluorides. The compound is usually found as a hygroscopic solid. The caesium ions form a cubic crystal system.

At the beginning of the 1980s, CsF has been investigated as a possible scintillation detector for time-of-flight positron tomography, but its temporal resolution was found to be unsuitable for that purpose.

**Caesium iodide**

*(Diagnostic Radiology)* Caesium iodide (CsI) is used in medical x-ray imaging in structured or columnar phosphor screens, for both flat panel detectors (indirect detection) and digital fluoroscopy. Within x-ray imaging, the phosphor screen is used to absorb the incident radiation (x-ray wavelength) and emit visible wavelength light photons, which are then detected through a number of optical imaging techniques. The wavelengths of the radiation absorbed and emitted by the phosphor is dependent on the material used in the phosphor screen, and the activator impurity introduced into it.

A structured phosphor has superior spatial resolution to that of the traditional powdered phosphor of the same thickness, as the
columnar structure acts like fibre optic light guide. The design of traditional powdered phosphor screens is restricted by a compromise between x-ray detection efficiency and spatial resolution (Figure C.1). As the thickness of the phosphor screen is increased, the probability of an incident x-ray interacting with it increases.

At the same time, by increasing the thickness, the emitted light photon must generally travel further from the point of emission before it exits the screen, increasing the spread of the produced light photons and thus reducing the spatial resolution. In contrast, the CsI columnar structure is produced by an evaporation process, which grows the phosphor to form columnar, crystalline structures. These crystals form an array of needle-like elements, which collimate the light and improve the spatial resolution. The spatial resolution is improved further by exposing the phosphor to a thermal shock which fractures the crystals allowing air to enter within the cracks, reducing the density by 15%–25%. The difference between the refractive indices of CsI and air (1.78 and 1, respectively) allows the crystals to act like fibre optics, which guide the light photons towards the horizontal edges of the phosphor with little lateral spread (assuming the photons are emitted at an angle that allows total internal reflection). The spatial resolution of a columnar screen is therefore less dependent on the screen thickness than a powder screen.

The k edges for CsI are at 33 and 36 keV, which gives a high absorption probability for the x-ray energies ordinarily used in radiography. In planar radiography, the CsI is generally activated by thallium (CsI:Th) that emits a green wavelength photon (∼550 nm), which is effectively absorbed in amorphous silicon photodiodes used in imaging detectors. In fluoroscopy, the CsI tends to be activated by sodium (CsI:Na) that emits blue wavelength light (∼450 nm), which is best matched to the response of photodiodes used in x-ray image intensifiers (XRII).

Although CsI has an improved spatial resolution compared to that of a powdered screen, it has several practical drawbacks as it is hydroscopic, toxic and very mechanically delicate. Due to these properties, it was first introduced as a phosphor for fluoroscopy since it could be protected within the image intensifier; however, recent advances in flat panel detectors have allowed its use in general radiography and mammography.

Related Article: Image intensifier

Caesium unit
(Radiotherapy) A Caesium unit is an obsolete machine in teletherapy that was used mainly in 1970s for palliative and non-tumourous treatments with short SSD.

Abbreviation: SSD = Source surface distance.
Related Article: Caesium

**Calcium**
(General)

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Ca</th>
</tr>
</thead>
<tbody>
<tr>
<td>Element category</td>
<td>Alkaline earth metal</td>
</tr>
<tr>
<td>Mass number A</td>
<td>40</td>
</tr>
<tr>
<td>Atomic number Z</td>
<td>20</td>
</tr>
<tr>
<td>Atomic weight</td>
<td>40.078 g/mol</td>
</tr>
<tr>
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<tr>
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</tr>
<tr>
<td>Boiling point</td>
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</tr>
<tr>
<td>Density near room temperature</td>
<td>1.378 g/cm³</td>
</tr>
</tbody>
</table>

**History:** Calcium is known to have been used in oxide form (lime) by the Ancient Romans. It was first isolated in its pure form by Sir Humphry Davy in 1808 using electrolysis of lime in mercuric oxide.

**Isotopes of Calcium:** There are four stable isotopes of calcium: ^{40}\text{Ca}, ^{42}\text{Ca}, ^{43}\text{Ca} and ^{44}\text{Ca}, and two metastable isotopes with mass numbers 46 and 48, which have half-lives sufficiently long to be considered stable. ^{40}\text{Ca} is the most abundant natural isotope, with a relative abundance of 97%. There also exists a radioactive cosmogenic isotope, ^{44}\text{Ca}, which is produced in the upper layers of soil by reaction of ^{40}\text{Ca} with neutrons and has a half-life of 103,000 years.
Medical Applications: X-ray Image Intensifiers – Calcium tungstate (CaWO₄) is a compound of calcium, which exhibits fluorescence under short-wave ultraviolet light. Large crystals of calcium tungstate were routinely used until the mid-1970s as fluorescent screens in real-time x-ray fluoroscopy, with a conversion efficiency of 5% and emission wavelength of 420nm. However, these have been superseded by rare earth metal screens, which have been found to be more suitable for the application.

Wound Dressings – Calcium alginate in powder form is used in modern wound dressings to absorb fluids secreted from wounds. On contact with fluid, it becomes a thick, non-adherent gel that provides a moist environment, which promotes tissue healing.

Calcium Scoring in CT – Calcium is present in hydroxyapatite, which makes up 70% of bone. It is also important in the nervous system and muscle contraction. Calcium is present in calcified plaques, which build up in the wall of the coronary artery and pose a major health concern. The disease can be diagnosed by measuring the amount of calcium present in the artery using a computed tomography (CT) scan, a process known as calcium scoring.

Related Articles: Image intensifier, CT (computed tomography)

Calculation of absorbed dose
(Radiotherapy) The dose can be measured either directly by a calorimeter observing the heating effect produced by the interactions of the radiation with matter or indirectly by ionisation chamber measurements under so-called Bragg–Gray conditions or by chemical dosimetry (e.g. Fricke dosimeter). These detectors are not suitable for quantifying the dose and its distribution in patients and therefore algorithms have been developed for calculating the dose distribution throughout the three-dimensional irradiated volume.

The core of the calculation process is a treatment planning system (TPS) that schematically comprises a computer, input and output devices and appropriate software. The TPS permits the input of the patient anatomical information and produces a representation of the dose distribution in the patient with as much accuracy as possible.

There are many types of dose calculation algorithms, but basically they can be classified into correction-based and model-based algorithms. In correction-based approaches, the dose is first measured in a water phantom using broad irradiation field or reconstructed from a representative sample of these measurements. Then the dose distribution is corrected for the presence of beam modifiers, tissue heterogeneities and patient surface contours. Model-based methods directly compute the dose to a patient, taking into account the beam energy and geometry, and the anatomical description of the patient. The most common methods for dose calculation are, respectively, the convolution/superposition and the Monte Carlo methods for computing photon beam dose and the Hogstrom pencil beam and the Voxel Monte Carlo methods for computing electron beam dose.

Related Articles: Treatment planning system, Convolution method, Hogstrom algorithm

Calibration
(Radiation Protection) Calibration of a measuring instrument (device) consists of establishing the relationship between its readings and the readings of another instrument known as the standard.

Calibration curve
(Radiation Protection) Calibration curve (Figure C.2) represents the relationship between a measured signal (dependent variable, e.g. number of counts) and the corresponding value of an independent physical variable, for example absorbed dose.

The linear regression method is used for fitting straight lines.

**Calibration factor**  
*(Radiation Protection)* The detector used for absorbed dose measurement in electron and photon beams in external beam radiotherapy should be calibrated using primary standards for absorbed dose to water. The calibration factors are measured for each ionisation chamber and beam quality in terms of exposure or air kerma according to the dosimetry protocol recommended by, for example IAEA (2000).

The measured dose $D$ expressed as a product of a calibration factor $k$ and the registered signal $M$ should be proportional to $M$:

$$D = k \times M$$

In reality, calibration factor $k$ depends on a number of factors, including the radiation (photons, electrons etc.), the energy (linear energy transfer (LET)) and intensity of the radiation; environmental parameters, for example temperature, pressure etc; and the characteristics of the detector, for example wall correction factor.

**Abbreviations**: IAEA = International Atomic Energy Agency and LET = Linear energy transfer.

**Related Articles**: Absorbed dose distribution, Calibration, Calibration curve, Calibration depth, Ionisation chamber, Primary standard


**Calibration source**  
*(Radiotherapy, Brachytherapy)*

**Calibration Source – Quality Control of Well-Type Chamber Stability**: The well-type ionisation chamber is commonly used for brachytherapy source calibrations, both at Secondary Standards Dosimetry Laboratories (SSDLs) and at hospitals. At the hospital level, it is important to have a chamber that is stable, reliable and easy to use, and the well-type chamber is the chamber recommended by the IAEA for photon-emitting sources (and for some beta sources).

The stability of any hospital ionisation chamber must be verified as part of the quality control of the chambers (recalibrations of the hospital standard chambers are recommended every 2 years). For the long-term stability/constancy checks of a well-type chamber, a source with long half-life, for example Cs-137 and an insert with stable geometry can be used. Alternatively, the stability checks can be made by irradiation in a Co-60 beam under stable geometric conditions.

**Calibration Source – Quality Control of Source Strength for ‘Permanent Seeds’**: The whole quality control procedure for an interactive permanent seed implant must be performed before the start of the actual implantation of the seeds themselves, including the verification of source strength. Source strength is verified using a well-type chamber calibrated for the type of source used and using the ‘dedicated insert’, specific for that type of source. In practice, a measurement of source strength is often made for a sample of at least 10% of the number of sources to be used for the implant (AAPM recommendation: 3% tolerance between hospital calibration and manufacturer source certificate for the mean of a batch of sources, and a maximum deviation of 5% for an individual source).

Verification of stability of the well-type chamber for the specific source type used can also be made using a single source, preferably with a smaller calibration uncertainty than the standard sources, ‘a calibration source’.

**Abbreviations**: AAPM = American Association of Physicists in Medicine and IAEA = International Atomic Energy Agency.

**Related Article**: Well-type ion chamber


**Calorimeter**  
*(Radiation Protection)* Calorimeter is a device measuring the absorbed dose by assessing the heating effect of the radiation (in Latin ‘calor’ means ‘heat’). Calorimeters are often made from graphite. The heat measurement is based on the fact that heat $\Delta Q_1$ lost by $B_1$ is equal to the amount $\Delta Q_2$ gained by $B_2$:

$$-\Delta Q_1 = \Delta Q_2 \rightarrow \Delta Q_1 + \Delta Q_2 = 0$$

The conservation law of heat enables to define the specific heat $c$, as the amount of heat $\Delta Q$ lost or gained by a body per unit mass, per unit temperature changes $\Delta T$:

$$\Delta Q = c \times m \times \Delta T$$

where $m$ is body mass.

According to the first law of thermodynamics (conservation of energy, including heat), the change in the internal energy $\Delta U$ of a system is equal to the heat $\Delta Q$ added to the system minus the work $W$ done by the system:

$$\Delta U = \Delta Q - W$$

When $W = 0 \rightarrow \Delta U = \Delta Q$.

This way the heat can be expressed in units of energy:

$$1 \text{ cal} = 4.186 \text{ J} \rightarrow 1 \text{ J} = 0.239 \text{ cal}$$

The amount of heat produced in the medium of known specific heat $c$, and mass, placed in an insulated container, can be measured calorimetrically as change of temperature ($\Delta T$). This measurement enables us to evaluate the absolute amount of energy absorbed by this medium.

**Related Article**: Calorimetry


**Calorimetry**  
*(Radiation Protection)* Calorimetry is a method of heat measurement in the medium, which attenuates x-rays or gamma radiation. The radiation x or gamma passing through the medium interacts with its atoms by producing the electrons (photoelectrons, recoil electrons of Compton scattering, pair production) and causes many ionisations. The kinetic energy of electrons is absorbed by the atoms of the medium, resulting in an increase in their internal energy $\Delta U$, which can be measured quantitatively (according to the first law of thermodynamics $\Delta U = \Delta Q$) as a temperature rise $\Delta T$. The produced heat $\Delta Q$ is equal:
\[ \Delta Q = c_s \times m \times \Delta T \]

where
\( c_s \) (J/kg/K) is the specific heat capacity of the medium
\( m \) (kg) is its mass
\( \Delta T \) (K) is measured temperature rise

Using calorimetry, one can perform an absolute measurement of absorbed dose \( D \) as a ratio of absorbed energy per unit mass of the body (medium):

\[ D = \frac{DQ}{m} \text{ (J/kg Gy)} \rightarrow D = c_s \times \Delta T \]

Example: for \( D = 1 \text{ Gy} \) \( \rightarrow \Delta T \) about \( 10^{-4} \text{ K} \)

This method is used for very large absorbed dose of radiation.

**Related Article:** Calorimeter


**CAP (computer-aided perception)**

*(General)* See Computer-aided perception (CAP)

**Capacitance**

*(General)* Capacitance is the property of body to store electrical charge. The symbol for capacitance is \( C \), and its value is expressed in farads (F). It is defined as the ratio of the charge \( Q \) (C) stored on the body and the voltage difference \( U \) (V) across the conductive parts of the body. The capacitance of a body is determined by the size and the shape of its conductive parts and the dielectric properties of the material between the conductive parts.

The more common term in daily use is a capacitor, which is an electrical (electronic) component having primarily capacitive properties within the field of its intended use.

**Related Article:** Capacitor

**Capacitive micromachined ultrasound transducer (cMUT)**

*(Ultrasound)* The basic principle for a capacitive transducer is a conducting backplate with a matrix of small cavities, Figure C.4. The cavities are sealed with a thin insulating membrane with a conducting top layer. The electrodes are biased with a DC voltage and an AC signal can either be applied to move the membranes or be generated by vibrating the membranes. This is due to the fact that the distance between the two conducting planes acts like a capacitor with altering capacitance (Farads). One transducer element is built up of a matrix of these cavities. Advanced IC fabrication processes are used to fabricate the structure of the small cavities, thin membranes and electrodes with submicron precision.

In recent years, the cMUT technology has emerged as an alternative to piezoelectric ultrasound transducers offering advantages such as wide bandwidth, ease of fabricating large arrays and potential for integration with supporting electronic circuits (see Khuri-Yakub et al., 2000).


**Capacitive reactance**

*(General)* The imaginary part of the complex impedance \( Z \) due to presence of capacitance in an electrical circuit is given by

\[ Z = R + iX_c \]

where
\( R \) is the resistance
\( X_c \) is the capacitive reactance
\( i \) is equal to \( \sqrt{-1} \)

The capacitive reactance is given by

\[ X_c = \frac{1}{\omega C} \]

where
\( \omega \) is the angular frequency
\( C \) is the capacitance

Capacitive reactance decreases with frequency.

Capacitive reactance causes voltage and current to become out of phase in AC circuits. In purely capacitive circuits, the current leads the voltage by the phase angle 90°, Figure C.5. Reactance is expressed in ohms (Ω).

**Related Articles:** Capacitance, AC current, AC voltage

**Capacitor**

*(General)* A capacitor is an electrical or electronic component that has dominantly capacitive properties. It consists of at least two conductors or semiconductors separated by an insulator. The conductive parts of a capacitor are described as electrodes or plates.

The value of the capacitance of a capacitor depends on the area of the electrodes, the distance between the electrodes and the dielectric properties of the insulating material between the electrodes. For a...
A parallel-plate capacitor, consisting of two parallel conductive plates, both having an area $S$ and separated by a distance $d$ filled with a material having a relative dielectric constant $\varepsilon_r$, the capacitance is equal to

$$C = \varepsilon_0 \varepsilon_r \frac{S}{d}$$

where $\varepsilon_0$ is the absolute dielectric constant (permittivity of free space) and it equals $8.854 \times 10^{-12}$ F/m. The dielectric material used in capacitors may be solid, liquid or gaseous. Capacitors used in electronic circuits usually have a solid dielectric material separating the conductive plates. Accordingly, they are differentiated as ceramic, foil, mica, electrolytic and other types of capacitors (Figure C.6).

Ceramic capacitors have ceramic as the dielectric material. Ceramic has high permittivity and therefore enables building large-capacity capacitors of small volume. Nominal values of ceramic capacitors capacity span from sub-picofarad to a few microfarad values. Ceramic capacitors with nominal values larger then a few nanofarads are often built as multilayer capacitors, Figure C.7.

Foil capacitors use metallised plastic foils to form a capacitor. The foils are wound in order to get large areas of electrodes, i.e. large capacitances. The electrical characteristics of foil capacitors are determined by the insulator foil properties. Commonly used plastic materials are polypropylene, polystyrene, polyester, polycarbonate and others. Paper may also be used as insulating material for capacitors (Figure C.8).

Mica capacitors are high quality and reliability capacitors due to high dielectric strength and high chemical stability of the insulator used – mica (a silicate mineral). Mica capacitors are used in high voltage and high frequency applications. They also have excellent temperature characteristics.

Electrolytic capacitors have the dielectric layer formed by electrolytic process. One electrode (anode) is made out of a metal conductor, aluminium or tantalum, while the other is formed using conducting electrolyte. Electrolytic capacitors have high capacitance per unit volume and also large leakage currents (Figure C.9).

Capacitors using liquid or gaseous dielectric materials are often used in transducer circuits. Variable capacitors (trimmer capacitors) use air as insulator and the nominal value of their capacitance is usually up to a few tenths of picofarads. The symbols for capacitors are shown in Figures C.10 and C.11. The unit for expressing the capacitance of a capacitor is farad (F). Since the capacitance of most capacitors built in electrical and electronic circuits is several orders of magnitude smaller than a farad, subunits of farad are in use: millifarads (mF), microfarads (μF), nanofarads (nF) and picofarads (pF).

Most capacitors have a fixed nominal value of their capacitance.

**Related Articles:** Capacitance, Circuit(s) electrical.
Capacitor discharge generator

(Diagnostic Radiology) This high-voltage generator (HVG) is used mainly for low-power mobile x-ray equipment. Its design includes a power high-voltage capacitor (or set of capacitors) – ~1 μF, connected after the HV rectifier (Figure C.12). During the charging period, the capacitors in the set are normally connected in parallel in order to ensure equal charge for each one. During the exposure (discharge), the capacitors are connected in series, so their summary voltage can ensure the necessary high voltage (kV). Often a grid-controlled x-ray tube is used with these HVGs. The kVp waveform follows the discharge – i.e. the kV decreases during the exposure (see article on Voltage waveform). The kV drop is an important parameter for this HVG. Normally, it is 1 kV/mAs (i.e. if an 80 kVp/30 mA s exposure is made, then the exposure begins at 80 kVp, but ends at 50 kVp). Before each exposure, the capacitors need approximately 10 s to charge. This HVG is lighter than the battery-powered HVG and has no special requirements to the mains electrical supply, but it cannot be used for powerful radiographs (e.g. of thick body parts or large patients).

Related Articles: High-voltage generator, Monoblock generator, High-voltage circuit, Voltage waveform

Capacity

(General) As a general term, this defines the maximum quantity of a substance that an object can accommodate. The capacity can usually be defined in terms of the maximum volume of fluid that can be accommodated.

In electronics, the capacity refers to the amount of electrical charge that can be stored within the device when a specific potential difference is applied across its electrical connections. Specific components, capacitors, are available with specified capacities in the range from $10^{-9}$ to $10^{-2}$ F (C/V).

Related Article: Capacitor

Capillary blockade imaging

(Nuclear Medicine) Capillary blockade refers to the process of using radioactive particles to investigate patients with a suspected embolism, for example ^99m^Tc-MAA (macro aggregated albumin) is used for regional lung perfusion studies.

When injected intravenously, the size of the radioactive particle used in these studies must be bigger than red blood cells to allow them to gather in the capillaries. By studying the accumulation of the radioactive particles in the capillaries, it is possible to draw conclusions regarding the relative blood flow in a specific organ or a specific part of an organ. The particles that gather in the capillaries cause a number of micro-embolisms but the number of particles typically injected is not enough to block more than a small fraction of the capillaries.
Capillary flow heterogeneity

(Magnetic Resonance) The heterogeneity of microvascular flows is known to be an important determinant of the efficacy of oxygen delivery to tissue. For this, the function \( h(t) \) is introduced to describe the distribution of transit times required by the different tracer molecules to pass through the capillary system (following an instantaneous tracer input). In particular, \( h(t) \) is the distribution of transit times. The impulse tissue residue function \( R(t) \) can be obtained experimentally, for example by dynamic susceptibility contrast MRI, and \( h(t) \) can then be calculated as the derivative of \( R(t) \) with respect to time \( t \), i.e. \( h(t) = -dR(t)/dt \). By applying the central volume theorem (i.e. flow equals volume divided by transit time) to each transit time of the distribution given by \( h(t) \), and assuming that all vascular paths have the same volume \( v \), the corresponding distribution \( w(f) \) of flow components \( f \) can be calculated. The application of vascular modelling presented by King et al. and Østergaard et al. provides the following relationship:

\[
w(f)df = h(t)dt \iff w(f) = \frac{h(t)}{f} dt \quad \text{(C.1)}
\]

Furthermore, the central volume theorem \( f = vt \) implies that \( df = -vf^2 = -tf \). Substituting \( df = -vf \) in Equation C.1 gives the following:

\[
w(f) = -\left(\frac{t}{f}\right)h(t) = \left(\frac{t}{f}\right)\frac{dR(t)}{dt} \quad \text{(C.2)}
\]

Finally, the capillary flow distribution \( w(f) \) is typically normalised to have unit mean flow and area.

Statistical comparisons of flow heterogeneity, represented by \( w(f) \), observed in normal tissue and in ischaemic regions in patients with stroke have indicated that the ischaemic regions tend to show a more homogeneous flow distribution. It has also been suggested that the region that shows abnormal flow heterogeneity in the acute phase after stroke correlates well with the final infarct volume as that the region that shows abnormal flow heterogeneity in the acute phase after stroke correlates well with the final infarct volume.

\[\text{History:} \quad \text{Carbon-14} \quad ^{14}\text{C} \quad \text{Nuclear Medicine}\]

\[\text{Element:} \quad \text{carbon} \]
\[\text{Isotopes:} \quad 2 < N < 16 \]
\[\text{Atomic number (Z):} \quad 6 \]
\[\text{Neutron number (N):} \quad 8 \]
\[\text{Symbol:} \quad ^{14}\text{C} \]
\[\text{Production:} \quad \text{Reactor} \]
\[\text{Daughter:} \quad ^{14}\text{N} \]
\[\text{Half-life:} \quad 5730 \text{ years} \]
\[\text{Decay mode:} \quad \beta^- - \text{decay} \]
\[\text{Radiation:} \quad \beta^- - 156.5\text{keV (max)} \quad 52\text{keV (mean)} \]
\[\text{Gamma energy:} \quad \text{none} \]
\[\text{Skin dose rate from} \quad 1 \text{MBq}: \quad 0\mu\text{Sv/h at} \quad 30\text{cm (point source)}; \quad 0\mu\text{Sv/h at} \quad 1\text{m (10mL glass vial)} \]
\[\text{Total absorption} \quad \text{in} \quad \text{air} \quad 0.32\text{mm in tissue (Lucite),} \quad 25\text{cm range} \]
\[\text{Biological half-life:} \quad \text{bone 12–40 days} \]
\[\text{Critical organ:} \quad \text{soft tissue/fat} \]
\[\text{ALL} \quad 50\text{mSv):} \quad 90\text{MBq} \]
\[\text{Absorbed dose:} \quad 0.095\text{mGy/MBq muscle} \]
\[\text{Effective dose:} \quad 0.024\text{mSv/MBq (ingestion)}; \quad 0.002\text{mSv/MBq (inhalation)} \]

\[\text{Carbon-14} \quad ^{14}\text{C} \]

(General)

<table>
<thead>
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<th>Symbol</th>
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<td>Element category</td>
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<td>Melting point</td>
<td>3925 K</td>
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<td>Boiling point</td>
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<tr>
<td>Density near room temperature</td>
<td>1900–2300 kg/m³ (1.9–2.3 g/cm³)</td>
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</tbody>
</table>
Carbon dioxide as contrast agent

Clinical Applications: $^{14}$C can be used for in vitro nuclear medicine. It is used to a small extent for, for example glycerol tri[1-$^{14}$C]oleate test for fat malabsorption in the small intestine. Other examples are $^{14}$C-inulin and $^{14}$C-triolein for measurement of the glomerular filtration rate (GFR) and $^{14}$C-urea for CO$_2$ breath tests (Helicobacter pylori infection).

In chemical and biological research, a wide range of $^{14}$C-labelled radiochemicals are used, examples are $^{14}$C-lipids, nucleic acids, steroids, amino acids, proteins, carbon monoxide, carbon dioxide, dopamine receptor ligands and various other different drugs in pharmaceutical research.


Carbon dioxide as contrast agent
(Diagnostic Radiology) Carbon dioxide (CO$_2$) gas can be injected into a blood vessel to produce contrast. Compared to iodine contrast media, CO$_2$ produces negative contrast because of its low density and atomic number. Some advantages of CO$_2$ as a contrast agent are its low toxicity and allergic effects and rapid absorption and dissipation after an injection.

Cardiac blood-pool imaging
(Nuclear Medicine) Cardiac blood-pool imaging is also called gated blood-pool imaging or multi-gated acquisition scan (MUGA). For more information, see Multi-gated acquisition scan (MUGA).

Cardiac cineangiography
(Diagnostic Radiology) See Cineangiography

Cardiac gating
(Magnetic Resonance) When performing cardiac MR, one must deal with the fact that the heart is beating. During a heart beat, the heart is translated, rotated and twisted. Blood is pumped between atria and ventricles. If the aim is to produce morphological images of the heart, it is desired to freeze this motion. If, on the other hand, functional information is desired, the goal is to faithfully depict the motion. For both these cases, a synchronisation of the imaging sequence to the heart rhythm is necessary (Figure C.13).

For synchronisation, a device is used to detect the $R$-peak of the ECG. For cardiac imaging, VCG (vector ECG) is most commonly used. If this is not applicable, a PPU (peripheral pulse unit) can be used instead. The devices produce a signal at the detection of an R-peak, and this signal is fed into the MR-scanner as a trigger to the acquisition.

There are different strategies in how to collect the data in respect to the $R$-trigger. The most basic strategy is cardiac gating. Trigger delay ($t_d$) and gate width ($t_w$) are defined, and data is acquired in the time period where the gate is open. This method is especially well-suited for morphological series, for example coronary artery and infarct (delayed enhancement) imaging. The gate is then open during late diastole, where the heart is not moving.

For functional imaging, one needs to collect data over as large a part of the cardiac cycle as possible. The two approaches are prospective and retrospective triggering. The prospective triggering starts data acquisition immediately after the $R$-peak is detected, and collects data for a predefined time. The acquisition then waits for the next trigger signal. Retrospective triggering acquires data continuously, and uses the trigger signal to keep track of when in the cardiac cycle a specific set of data was measured. Retrospective triggering is the most commonly used method as it covers the complete RR-interval, whereas when using prospective triggering, information is not acquired in late diastole (Figure C.14).

Cardiac gating
(Nuclear Medicine) Cardiac gating is an imaging technique in which images are synchronised to electrocardiogram (ECG) signals, thus permitting images of the heart to be formed in different phases of the cardiac cycle. Cardiac images acquired without gating techniques are often blurred due to the contraction and relaxation of the cardiac muscle. The technique is used whenever the data acquisition is too slow to occur under a fraction of the cardiac cycle. The trigger position in the cardiac cycle is determined using the motion. It is sent to the reconstruction with a time stamp, indentifying from which phase in the cardiac cycle it belongs to.
ECG, the signal from which is fed live to the acquisition electronics. Images are thus acquired at different times in the cardiac cycle and arranged in a movie loop.

The typical count rates in nuclear medicine are too low to produce a sufficient image from a single cardiac cycle. Therefore, an image is typically comprised of an average of 50–100 images. A common diagnostic examination in nuclear medicine is gated blood-pool scanning (also known as multi-gated acquisition or MUGA). This type of scan can be used to determine the size of the ventricles, look for abnormalities in the heart wall (like aneurysms) and studying abnormal blood flow between the ventricles. It is used to assess the function of the heart by the calculation of the left ventricle ejection fraction.

**Related Article:** Multi-gated acquisition (MUGA)

**Cardiolite**  
*Nuclear Medicine* Cardiolite is a trade name for Technetium-99m Sestamibi. For further information see Tc-99m SestaMIBI

**Related Article:** Tc-99m SestaMIBI

**C-arm in fluoroscopy**  
*Diagnostic Radiology* Special type of C-shaped support holding the x-ray tube and image intensifier (II). The C-arm can be rotated around the patient around three rotational axes, thus allowing fluoroscopy from various angles. C-arm mobile fluoroscopic systems (Figure C.15) are very useful for intra-operative imaging, as these allow easy operational access of the personnel in the theatre. C-arm systems are also used for angiographic systems (sometimes using a special type of C-shaped support holding the x-ray tube and II is usually between 90 and 120°). Moving the II close or away from the tube.

**Related Articles:** Mobile unit, Fluoroscopy, Biplane cine system

**Carr–Purcell (CP)**  
*Magnetic Resonance* The Carr–Purcell pulse sequence consists of a 90° excitation RF pulse followed by a train of 180° RF refocusing pulses in order to recall several echoes after one excitation pulse. The time separation between the 90° pulse and the first 180° pulse is τ, and the separation between following 180° pulses is 2τ.

In the Carr–Purcell sequence, the phase angle of all RF pulses are the same, i.e. 90°, −τ− 180°, −2τ− 180°, −3τ− 180°, ..., which leads to an accumulation of phase errors and an additional reduction of the echo amplitude as the echo train is prolonged, occurring due to, for example imperfections of the 180° pulse. This limiting factor was solved in the CPMG pulse sequence, where a 90° phase shift was introduced between the excitation pulse and the successive refocusing pulses. Alternatively, a 180° phase cycling between successive 180° RF pulses also compensates for this problem.

The Carr–Purcell pulse sequence was originally created to measure the complete echo decay by creation of multiple recalled echoes needed, for example for T₂ measurements. This is also the first multi-echo pulse sequence.

**Related Article:** Carr–Purcell–Meiboom–Gill (CPMG) sequence

**Carr–Purcell–Meiboom–Gill (CPMG) sequence**  
*Magnetic Resonance* The Carr–Purcell–Meiboom–Gill pulse sequence is a modification of the Carr–Purcell pulse sequence. It consists of a 90° excitation RF pulse followed by a train of 180° RF refocusing pulses, which in contrast to the CP pulse sequence are phase shifted by 180°. The timing between the RF pulses is the same as in the CP pulse sequence, i.e. τ between the 90° and the 180° RF pulses and 2τ between successive 180° RF pulses. The CPMG pulse sequence hence consists of the following pulses; 90°, −τ− 180°, −2τ− 180°, −3τ− 180°, ..., the phase shift between successive 180° was introduced to inhibit the accumulation of phase errors.

**Related Article:** Carr–Purcell (CP)

**Carrier-added radioisotope**  
*Nuclear Medicine* When a known amount of the corresponding stable isotope has been added to the radioactive sample, it is referred to as carrier-added. The stable isotope is called the carrier.

**Carrier-free sample**  
*Nuclear Medicine* A sample with a radioisotope can also contain other isotopes of the same element. For instance, a ¹²¹I sample can also contain the stable isotope ¹²⁷I. When an isotope of the same element as the radionuclide of interest is present in a sample, the sample is said to be with carrier. A sample without a stable isotope is said to be carrier free. Such pure samples are pretty much impossible to manufacture without a small fraction of carrier atoms. Therefore, the term without carriers is also used to describe carrier-free solutions. If the carrier atoms are also radioactive, they are unlikely to share the same decay characteristics, i.e. one isotope is suitable for imaging while the other is not. The second isotope will only contribute to an increase in the total patient radiation dose without giving any diagnostic information. One always strives to obtain carrier-free samples, and the choice of production method is often decided by which of the methods produces the sample with the smallest amount of carriers. The sample specific activity is the ratio between radioisotope activity and the total mass of the element present. When there are no carrier isotopes of the same element as the radioisotope of interest, the specific activity is called carrier-free specific activity.


**Carrier-free specific activity (CFSA)**  
*Nuclear Medicine* Carrier-free specific activity (CFSA) is the highest possible specific activity of a radionuclide. A sample with a radioisotope can also contain stable isotopes of the same element.

![Figure C.15](image-url)  
**Figure C.15** Typical mobile fluoroscopic x-ray equipment with C-arm. In this system, the II (on top) is fixed.
For instance, a $^{131}$I sample can also contain the stable isotope $^{127}$I. When a stable isotope of the same element as the radionuclide of interest is present in a sample, the sample is said to be with carrier. A sample without a stable isotope is said to be carrier free. Such pure samples are pretty much impossible to manufacture without a small fraction of impurity atoms. Therefore, the term without carriers is also used to describe carrier-free solutions.

The sample specific activity is the ratio between radioisotope activity and the total mass of the element present. When there are no polluting isotopes of the same element as the radioisotope of interest, the specific activity is called carrier-free specific activity.


**Carrier-mediated diffusion of tracers**

*(Nuclear Medicine)* The term ‘carrier-mediated diffusion of tracers’ refers to a passive transport of tracers in the same direction as a concentration gradient within the body. An example of carrier-mediated transport is the transport of glucose and amino acids across the blood–brain barrier. The transport uses a carrier molecule to transport the substrate across a barrier. They form a substrate/carrier complex that can physically move across the barrier. At the other side of the barrier, the complex decays into the original substrate and carrier molecule. Since there are only a finite number of carrier molecules, this type of transport can be saturated. The carrier molecule is typically an enzyme that is neither created nor destroyed in the process but rather acts as a catalyst.

**Further Reading:** Cherry, S. R., J. A. Sorenson and M. E. Phelps. 2012. Physics in Nuclear Medicine, Saunders, Philadelphia, PA.

**Cartesian coordinates**

*(General)* The Cartesian coordinate system is a 2D reference system where every point can be determined by two coordinates. The coordinates are defined by two perpendicular lines with a unit length. The lines are generally called the x- and y-axis. The Cartesian coordinates can be used to model geometric shapes described by algebraic functions.

The 3D case of the Cartesian coordinate system provides the three physical dimensions of space – length, width and height. The location of a specific point in 3D Cartesian coordinate system is given by the x-, y- and z- coordinates.

**Cassette carriage**

*(Diagnostic Radiology)* A tray holding the cassette with x-ray film in radiography systems. There are two main types of carriage – used in patient table (see Undertable cassette carriage) or used in vertical x-ray stands for chest imaging. These are most often mounted in one assembly with the anti-scatter grid and the detectors (dominants, cells) of the automatic exposure control (AEC) system. The cassette carriage has metal clips holding different cassette sizes. The centre of the cassette carriage is aligned with the centre of the x-ray beam.

**Related Articles:** Radiography, Bucky table, Undertable cassette carriage

**Cassette changer**

*(Diagnostic Radiology)* Equipment used to quickly change the cassettes with x-ray films in x-ray radiography systems performing sequences of exposures. This equipment usually has a magazine loaded with cassettes with unexposed x-ray films; a mechanism quickly moving the cassette in front of the x-ray beam; and a magazine for the cassettes with exposed films.

Cassette changers used in fluoroscopic systems can perform up to – two to three exposures per second, while fast cassette changers used in angiography can have around 10 exposures per second. These systems are not used anymore, as contemporary digital x-ray equipment can acquire images with speed above 25 fps.

**Cassette, filmless**

*(Diagnostic Radiology)* An old name for computed radiography (CR) using storage phosphor plate (the cassette does not contain a film, but a storage phosphor plate).

**Related Article:** Computed radiography

**Cassette size**

*(Diagnostic Radiology)* The sizes of x-ray films and their holders (x-ray cassettes) are standardised according to the anatomical regions to be radiographed. Usually, the sizes are two types – American/English using inches (e.g. 8 × 10; 10 × 12 in., etc.), and European using centimetres (e.g. 18 × 24, 24 × 30 cm, etc.).

**Cassette, Wisconsin**

*(Diagnostic Radiology)* See Wisconsin test cassette

**Catapult bucky**

*(Diagnostic Radiology)* A mechanism that moves the anti-scatter grid during the x-ray exposure to blur and reduce the visibility of the grid lines in the image. It is also known as a reciprocating Bucky (or in a specific company device – Lysholm raster).

**Cathode (of an x-ray tube)**

*(Diagnostic Radiology)* The cathode is the negatively charged source of electrons in the x-ray tube. In almost all contemporary tubes, the cathode assembly (Figure C.16) consists of a heated tungsten wire (filament), which produces thermal electrons.

**FIGURE C.16** Cathode of an x-ray tube: (1) filament of the broad focus (in its focusing cup); (2) filament of the fine focus (in its focusing cup) and (3) cathode side of the x-ray tube (photo from a broken x-ray tube). (Graphs courtesy of EMERALD project, www.emerald2.eu)
Cathode ray tube

The emission of thermal electrons depends greatly on the temperature of the cathode. In order to obtain an anode current on the order of several amperes (for heavy x-ray exposure), the cathode must be heated to some 2000–2700 K. Due to this reason, the material used for preparation of x-ray cathodes is normally tungsten. Its high melting point (3410°C) and its ability to be drawn into very thin wire were the most important parameters to be considered for this choice. Normally, the cathode is a long (~5–20 mm), thin (~0.1–0.3 mm) tungsten wire in spiral coil with diameter on the order of 0.2 mm. One of the ends of the filament is also connected to the negative side of the high voltage. The electrical resistance of cathode filament is relatively high and changes from approximately 0.1–0.3 Ω when cold, to some 2–6 Ω when heated above 2000 K (ohmic heating).

The very high temperature of the filament leads to some evaporation of the tungsten wire. This evaporation leads to shortening of the life of the cathode (thinning it). Normally, the cathode life at this temperature is not more than 1000 working hours. Due to this reason, the cathode is heated to this high temperature for limited time only (during the x-ray exposure). But to heat the cathode from room temperature to 2700 K takes time. In order to keep the heating time short, the cathode always stays preheated to temperature around 1500 K. The preheating is made by applying a constant stand-by filament current through the cathode (less than 1 A).

The variation of the anode current (the thermal electrons flying from cathode to anode) is achieved by changing the temperature of the cathode, which in turn is achieved by changing the filament current (I_f). The density of the thermal emission current is described with the Richardson equation:

\[ J_0 = A_0 \cdot T^2 \cdot e^{-w/kT} \]

where

- \( J_0 \) is density of the emission current
- \( T \) is temperature of the emitter (in Kelvin degrees)
- \( k \) and \( w \) are constants (\( k = \text{Boltzmann constant}, \ w = \text{work function, for tungsten} = 4.5 \text{eV} \))
- \( A_0 \) is constant depending on the material of the emitter (for tungsten = 60 A/cm²/K²)

The electrons emitted from the heated cathode form an electron cloud around it. This cloud is called ‘space charge’. When the electrons leave the filament, the cathode loses part of its negative charge and becomes more ‘positive’. This attracts back some of the electrons. Normally, an equilibrium state exists between the number of electrons emitted and those attracted back. The cloud remains near the filament until high voltage (from 20–150 kV) is applied between the cathode and the anode. Thus, the thermal electrons are accelerated towards the anode by the electric field between the cathode and anode, forming the anode current. The area of the anode, which is bombarded by the thermal electrons and produces x-rays, is called actual (or thermal) focal spot. The size of the actual focal spot depends on the size of the cathode wire (in fact, the size of the electron beam generated by it). Most of the x-ray tubes have two foci – one small filament wire (up to 1 mm long) – known as ‘fine focus’ and one bigger filament wire (~2–3 mm long) – known as the ‘broad focus’. The foci (the filament coils) can be placed side by side (as in the Figure C.16); in a line (one above the other); in different depths (one behind the other), etc. (there are various designs).

Normally, the beam of thermal electrons produced by the cathode filament is quite spread, resulting in an increased area of the focal spot. This enlarged size of the source of radiation blurs the x-ray image. In order to focus the beam of thermal electrons and to decrease the space charge effect, the cathode filament is placed in a focusing cup (a half-pipe groove, known also as Wehnelt electrode, or Wehnelt cylinder). The focusing cap is specially shaped and is made of molybdenum, nickel or steel, because of their poor thermionic emission. The cup can be equipotential with the cathode, or under negative potential (Wehnelt electrode, used in grid control x-ray tubes).

Related Articles: Filament circuit, Filament heating, Wehnelt electrode, X-ray tube, Focal spot, Stationary anode, Rotation anode

Cathode ray tube (Diagnostic Radiology) A cathode ray tube (CRT) is an evacuated electronic tube (in glass envelope) with a cathode (electron source) at one end and a fluorescent screen at the other. The electrons from the cathode are focused into a small beam and accelerated to the fluorescent screen by a set of electrodes, including an anode operating at a high voltage relative to the cathode.

A bright spot is produced where the electron beam strikes the fluorescent screen.

The beam can be deflected and the bright spot moved over the screen surface either by a potential applied to electrodes within the tube or currents applied to magnetic coils located around the tube.

A CRT is the component used in an oscilloscope to display various waveforms applied to deflect the beam, usually in the vertical direction, while the beam is being scanned in the horizontal direction.

CRTs are used for the display of video (television) images. The beam is scanned over the surface of the screen in a pattern of horizontal lines that are progressively moved in the vertical direction to cover the image area.

At each beam location within the image, the intensity of the electron beam is modulated by the video signal transmitted from the camera to control the brightness of each spot in the displayed image.

These days the diagnostic TV monitors with CRT are being replaced by flat-panel displays.

Hyperlink: The cathode ray tube site: http://members.chello.nl/~h.dijkstra19/page3.html

Cathode rays (Diagnostic Radiology) Cathode rays are electrons emitted from a cathode in an evacuated tube such as an x-ray tube or a cathode ray (CR) tube.

Related Article: Cathode ray tube


Cavitation (Ultrasound) Cavitation is a general term that covers a wide range of phenomena all having as a common factor that a cavity is formed within a liquid, whether the cavity is empty, or containing gas or vapour. Formation of cavities may be, for example as a result of explosions or boiling. Acoustic cavitation is, in the context of medical ultrasound, formation of a cavity in response to an acoustic field. Such cavities are formed from pre-existing microscopic bubbles or bubble nuclei with a size on the order of microns or also from impurities in the liquid that can act as a seed for cavitation.

There are two distinct categories of cavitation. In stable cavitation, a bubble undergoes periodic pulsations in an acoustic field. In transient cavitation, which occurs at higher acoustic pressures, bubbles reach a maximum size and then rapidly collapse. The rapid...
collapse is a high-energy event and carry potentially destructive effects such as, for instance, breaking cell membranes. The sound energy (sound pressure) needs to be above a certain threshold level for a bubble to go from stable to transient cavitation. In order to give ultrasound operators an indication of the degree of possible risk of transient cavitation, the mechanical index has been introduced. It should be noted that this index only gives indication of the likelihood of cavitation and the effect responsible for tissue damage may be some other mechanism. Further, the model for this index assumes a spherical nucleus, and other effects may occur for other shapes of nuclei.

**Cavity-gas calibration factor**

(Radiotherapy) The cavity-gas calibration factor indicated by \( N_{\text{gas}} \) is the dose to the gas in the ionisation chamber per unit electrometer reading. It is constant for all the radiation qualities for which the average energy expended in the production of one ion pair (\( W \)) is the same as that for Co-60 gamma rays (33.7 eV). The absorbed dose to the air in the cavity of an ionisation chamber is related to the dose to the wall through the Bragg–Gray equation and the dose to the wall material can be related to the dose to air in absence of the ionisation chamber, which is also related to the exposure assuming a condition of electronic equilibrium. Consequently, dose to the air in the cavity is related to the exposure calibration factor of an ionisation chamber.

\[
D_{\text{gas}} = D_{\text{ion}} \times \left( \frac{W}{e} \right)
\]

where

- \( D_{\text{gas}} \) is the dose to the gas (Gy)
- \( W/e \) is the quotient of the average energy expended to produce an ion pair divided by the electronic charge; for room air, \( W/e = 33.7 \text{ J/C} \)
- \( D_{\text{ion}} \) is assumed to be corrected for ion recombination

As the response of the electrometer is also directly related to \( D_{\text{gas}} \), the quotient of \( D_{\text{gas}} \) by the electrometer reading \( M \) is a constant, which depends on the dimensions and composition of the ionisation chamber. The ratio \( D_{\text{gas}}/M \) is referred to as cavity-gas calibration factor \( N_{\text{gas}} \), i.e.

\[
N_{\text{gas}} = D_{\text{gas}} \frac{A_{\text{ion}}}{M}
\]

where

- \( A_{\text{ion}} \) is the ionisation collection efficiency at the time of calibration at a primary dosimetry laboratory
- \( N_{\text{gas}} \) is unique to each ionisation chamber and does not depend on the composition of the dosimetry phantom. It is applicable to all ionising radiations for which \( W/e \) has the value quoted earlier.

\( N_{\text{gas}} \) can be calculated using \( N_{x} \), exposure calibration factor (R/C or R/scale division), uncorrected for ion recombination from

\[
N_{\text{gas}} = N_{x} \frac{k(W/e)A_{\text{ion}}A_{\text{wall}}\beta_{\text{wall}}}{(E/p)_{\text{gas}}(\mu_{\text{eff}}/\rho)_{\text{wall}}}
\]

where

- \( A_{\text{wall}} \) is a factor that corrects for attenuation and scatter in the wall and build-up cap
- \( A_{\text{ion}} \) is the ion collection efficiency in the user’s chamber at the time of Co60 exposure calibration at a primary dosimetry laboratory
- \( k \) is the charge produced in air per unit mass per unit exposure \((2.58 \times 10^{-4} \text{ C/kg/R})\)
- \( \beta_{\text{wall}} \) is the quotient of absorbed dose divided by the collision fraction of kerma in the chamber wall, 1.005
- \( (E/p)_{\text{gas}} \) is the ratio of the mean restricted collision mass stopping power of the wall material to that of the gas
- \( (\mu_{\text{eff}}/\rho)_{\text{wall}} \) is the ratio of the mean mass energy absorption coefficient for air to that of the wall for Co-60 gamma rays

**Calcium tungstate (CaWO₄)**

(Diagnostic Radiology) CaWO₄ (crystalline calcium tungstate) is the original phosphor used for x-ray intensifying screens (since 1896). This material can be found naturally (scheelite), but it is its synthetic form that is used for radiographic screen – films. Calcium tungstate must be free of impurities to produce good fluorescence. It emits light from 350 to 580 nm (with peak at 430 nm). This violet light is not seen well by human eye, but the photographic emulsion of specific radiographic films is very sensitive to it. Calcium tungstate has conversion efficiency on the order of 15 (while CsI has ~45 and CdWO₄ has ~40) and is now replaced by new screens using various rare earth elements.

**Related Article:** Screen film


**CBF (cerebral blood flow)**

(Magnetic Resonance) See Cerebral blood flow (CBF)

**CBV (cerebral blood volume)**

(Magnetic Resonance) See Cerebral blood volume (CBV)

**Ceilng mount unit**

(Diagnostic Radiology) x-ray stand mounted on the ceiling (usually without the generator). For simple radiography units, the ceiling unit holds only the x-ray tube on a telescopic column. Ceiling-mounted complex C-arm units (used mainly in angiographic devices) hold the tube plus the image intensifier.

**Related Article:** C-arm

**Cell cycle**

(Radiotherapy) The cell cycle is the series of events that take place in a cell, leading to its replication, and it is defined as the interval between the midpoint of mitosis in a cell and the midpoint of the subsequent mitosis in both its daughter cells (mitosis is the process of cell division resulting in the production of two daughter cells from a single parent cell. It is divided into five sub-phases: prophase, prometaphase, metaphase, anaphase and telophase). The cell cycle follows an orderly sequence of phases between one mitosis (\( M \) phase) to the next. These phases are the S phase, during which DNA synthesis occurs, and the \( G_1 \) and \( G_2 \) phases, gaps before and after the \( S \) phase, respectively. Additionally after mitosis, cells may spend some time in a ‘resting’ phase \( G_0 \).

A set of cyclin-dependent kinases (CDK) have a major role in the regulation of the cell cycle and act via a series of cell cycle
checkpoints set at various stages of the cell cycle. If DNA damage is detected, for example the checkpoint will either delay the cell cycle until the damage is repaired or, if repair is not possible, target the cell for destruction via programmed cell death (apoptosis, also sometimes referred to as cell suicide).

**Cell proliferation**  
(*Radiotherapy*) Cell proliferation refers to the increase in the number of cells as a result of growth and cell division. Cancer cells demonstrate loss of proliferative control, whereby a group of cells display uncontrolled growth.  
**Related Article:** Cell cycle

**Cell survival**  
(*Radiotherapy*) Cell survival, and its opposite cell death, have different meanings depending on the context. For non-proliferating cells such as nerve and muscle, death can be defined as the loss of a specific function. For proliferating cells such as stem cells, the appropriate definition is the loss of capacity for sustained proliferation. This is sometimes called reproductive death and is the end point measured with cells cultured in vitro. A survivor that has retained its ability to reproduce and thus can proliferate indefinitely to produce a large clone or colony is said to be clonogenic.  
**Related Articles:** Cell cycle, Cell proliferation  

**Cell survival curve**  
(*Radiotherapy*) A plot of surviving fraction against radiation dose is called a *cell survival curve* and is usually presented in the form shown in Figure C.17 for x- and γ-rays, with dose plotted on a linear scale and surviving fraction on a logarithmic scale.  
A cell's radiosensitivity can be determined from cell survival curves. The shape of survival curves is commonly described by the linear-quadratic model with the ratio of its two parameters, α and β, often used as a measure of a cell's radiosensitivity.  
For further information, see the article on *Linear-quadratic (LQ) model*.  
**Related Articles:** Alpha–beta ratio, Cell survival, Dose response model, Linear-quadratic (LQ) model, Probability of cell survival, Radiosensitivity

**Central beam**  
(*Diagnostic Radiology*) See *Central ray of x-ray beam*

**Central field of view (CFOV)**  
(*Nuclear Medicine*) An image system is able to depict an area or a volume of a certain size. This size is referred to as the image systems field of view, FOV. Due to certain system properties, information close to the edges is not suitable for imaging or activation quantification. The portion of the FOV used for these purposes is called useful field of view, UFOV. During the quality control of the camera system, the resolution, linearity and uniformity are measured over the UFOV and also over the central field of view, CFOV. The CFOV is defined as 75% of UFOV and is the smallest of the three FOVs.  
**Related Articles:** Field of view (FOV), PET, SPECT, Useful field of view

**Central processing unit (CPU)**  
(*General*) The central processing unit (CPU) is the computing part of a computer system, also called processor. Its main components are the arithmetic logic unit (ALU) and a control unit. Usually, the CPU is contained in a single chip. Today, all CPUs are microprocessors (MPU) – a CPU miniaturised in a single chip. Some medical imaging equipment has a special imaging processor (Imager), which is an additional CPU used specifically for the acquisition and processing of images.  
**Hyperlink:** [www.pcmag.com/encyclopedia_term/](http://www.pcmag.com/encyclopedia_term/)

**Central ray of x-ray beam**  
(*Diagnostic Radiology*) The central ray of an x-ray beam (also known as central beam) is the ray with maximal intensity focussed at the centre of the detector (x-ray beam). Normally, this ray is perpendicular to the cathode–anode line.  
See the article on Anode heel effect (Figure A.45).  
**Related Article:** Anode heel effect

**Central volume theorem**  
(*Nuclear Medicine*) The central volume theorem is used to calculate the cerebral blood flow (CBF) from the relation CBF equals the cerebral blood volume (CBV) divided by the mean transit time (MTT), or

\[
\text{CBF} = \frac{\text{CBV}}{\text{MTT}}
\]
The MTT is the average time the radiopharmaceutical takes to pass through the tissue being studied. Cerebral blood volume (CBV) and blood flow (CBF) are parameters derived from the first pass of the radiopharmaceutical.

**Centre of rotation**

(Nuclear Medicine) The centre of rotation (COR) is a stationary point at the centre of a tomographic imaging system. In a SPECT imaging system, this point correlates to the midpoint of the detectors circular orbit around the patient. COR should coincide with the mechanical centre of rotation; any deviations could induce image artefacts, for example ring artefacts or additional blurring.

In a PET system, the COR is the same as the midpoint of the detector circle. A part of the camera system routine quality control is measuring and correcting for any deviation of COR from the mechanical centre of rotation.


**Related Articles:** PET, SPECT

**Centric sampling**

(Magnetic Resonance) Centric sampling refers to the order in which lines of k-space are acquired. The centric sampling scheme can be applied in the phase-encoding direction for 2D pulse sequences and in the phase- or slice-encoding direction for 3D pulse sequences. For 3D, although many variants exist, the filling of k-space starts at the centre (k = 0), typically followed by adjacent lines in an interleaved manner (k = −1, 1, −2, 2,...) as shown in Figure C.18 (slice encoding example).

Centric sampling is applied when the contrast is changing during the k-space loop, so that conventional linear sampling would result in a loss of contrast and/or SNR.

Different sampling patterns can be used in contrast-enhanced angiography applications in order to control the sampling of the k-space relative to the arterial (early) and venous (late) phases of the contrast agent bolus. Centric sampling results in early sampling of the k-space centre, which means that the image contrast is highly dependent on the amount and location of contrast agent at the beginning of image acquisition. Sampling of k-space in combination with the passage of the contrast agent also acts as a low-pass filter, which gives some blurring of the image of the vessels along different directions depending on the sampling pattern.

When a 3D sequence with two phase-encoding directions (k and k) is used for contrast-enhanced angiography, only the outer slower loop (corresponding to TR) is normally considered for centric sampling. Centric sampling of the faster inner loop can in some applications be used to control fat saturation. It is also increasingly used in T1-weighted magnetisation-prepared 3D sequences.

**Related Articles:** Bolus, Contrast-enhanced angiography, k-space

**Ceramic capacitor**

(General) See Capacitor

**Ceramic x-ray tube with double bearings**

(Diagnostic Radiology) The first ceramic x-ray tube with double bearings was developed by PHILIPS (Super-Rotalix-Ceramic tube). This rotating anode tube has an unusual design – the anode disc is supported on both sides (Figure C.19). This way the anode stem has two bearings (one at each end), which makes the rotation more stable (in comparison with ordinary x-ray tubes where there is only one bearing at the anode stem – under the rotor, behind the anode). The x-ray tube with double bearings is very powerful – i.e. the anode temperature is high and good cooling is necessary. This requires a metal tube housing with internal ceramic coating. This tube has three high-voltage ceramic insulators (aluminium oxide) – two for the high-voltage cables for the anode and cathode (both at one side of the tube) and a third one at the anode stem behind the anode (under the rotor winding).

**Related Articles:** Anode, Rotating anode, Metal x-ray tube, Bearing

**Ceramics**

(General) The word ceramic may be used as a noun or an adjective, referring to inorganic and non-metallic materials formed by the solidification of a molten substance with the action of heat. Ceramic materials tend to be hard, porous and brittle, and they have a crystalline or amorphous structure. Non-crystalline ceramics, such as glasses, tend to be formed from melts and are shaped by casting or blowing. Crystalline ceramic materials are made by reaction in situ, such as cement and concrete, or by sintering shaped powders to form a solid body, such as pottery.
Ceramics can be categorised as structural (e.g. bricks and roof tiles), refractory (e.g. kiln linings and gas fire radiants), whiteware (e.g. sanitary ware, tableware and wall tiles) and technical (e.g. gas burner nozzles, biomedical implants and jet engine turbine blades). Technical ceramics can be further classified into the following: oxides, such as alumina; non-oxides, such as silicides; and composites, which are particulate reinforced combinations of oxides and non-oxides.

Ceramics, such as transition metal oxides, can be semiconductors, which are useful for surge protection applications and gas sensors. Some ceramics display superconductivity at extremely low temperatures or 'high' temperatures (>30 K). Other ceramics, such as quartz, exhibit piezoelectricity, the generation of an electric potential when under mechanical stress. Piezoelectric ceramics are used for accurate measurements, as crystal oscillators in electronic circuits of watches and computers, and as sensors including ultrasound transducers. Piezoelectric materials can also be pyroelectric, meaning that they produce an electric potential with changes in temperature, for use as motion sensors. In addition, pyroelectric materials can also be ferroelectric, meaning that an electric dipole can be oriented or reversed by applying an electrostatic field. This property can be used to store information in ferroelectric capacitors.

Ceramics have a wide range of applications due to their specific properties. They are used for knife blades as they remain sharper for longer than steel blades; however, they are more likely to break because they are brittle. Some ceramics, such as alumina, are used in armoured vests and military airplanes, due to their low weight. Ceramic balls often replace steel in ball bearings as they are harder, experience less deformation, build up less heat, are more chemically resistant and electrically insulating. However, the main disadvantage of ceramics is that they are more expensive. Ceramics can be used in engines, as they do not require a cooling system, leading to greater fuel efficiency although they are not widely used in engines as precision and durability are difficult to achieve and the high degree of imperfections lead to a high risk of equipment failure. Ceramics are also used for producing watch cases, as they are lightweight, scratch-resistant and durable.

**Medical Applications:** Ceramics are widely used in medicine, for example bio-ceramics for dental implants and synthetic bone substitutes. Hydroxyapatite, the mineral in bone, can be made synthetically and formed into ceramic materials. They are used to make orthopaedic implants that bond readily to body tissues without rejection or inflammation. Such ceramics are often porous, lacking strength, and so usually coat metal orthopaedic implants or are used as bone fillers. They are being developed for gene delivery and tissue engineering scaffolds.

Ceramic piezoelectric sensors, typically lead zirconate titanate (PZT), are used in ultrasonic transducers for medical imaging. Ultrasound transducers are also used in physiotherapy, lithotripsy to obliterate kidney stones and HIFU (high-intensity focused ultrasound) for destruction of pathogenic tissue.

**Related Articles:** Quartz. High-intensity focused ultrasound (HIFU), PZT, Transducer

**Cerebral blood flow (CBF)** *(Magnetic Resonance)* The cerebral blood flow (CBF) is the rate at which blood passes through the capillary network in brain tissue, typically expressed as volume of blood per unit of time. Traditionally, this quantity is also known as blood perfusion. A common unit for regional CBF (often denoted rCBF) is mL/(min 100 g), i.e. millilitres of blood per minute per 100 g of tissue. Through exchange processes, taking place at the interface between the capillary system and tissue, the blood carries oxygen and nutrients to the brain tissue and removes carbon dioxide and other waste products. Hence, the CBF is extremely important for maintaining tissue viability. According to Leenders et al. (1990), the CBF of normal grey matter is on the order of 50–60 mL/(min 100 g) and the CBF of normal white matter is on the order of 20–25 mL/(min 100 g). MRI-based methods for assessment of CBF are dynamic susceptibility contrast MRI (DSC-MRI) and arterial spin labelling (ASL).

The central volume theorem states that the CBF is related to the cerebral blood volume (CBV) and the mean transit time (MTT) according to the following expression:

\[
\text{CBF} = \frac{\text{CBV}}{\text{MTT}}
\]

The determination of the CBV and CBF is based on the principles of tracer kinetics for non-diffusible tracers Meier and Zierler (1954), Zierler (1962) and relies on the assumption that in the presence of an intact BBB, the contrast material remains intravascular. When such an intravascular tracer is employed (as in DSC-MRI), the CBV can be derived from the time integral of the tissue concentration-versus-time curve, normalised to the time integral of the corresponding arterial concentration-versus-time curve. According to Zierler's area-to-height relation, the MTT can be calculated as the ratio of the time integral of the tissue residue function \(R(t)\) (i.e. the fraction of tracer that remains in the tissue at a time \(t\) following an instantaneous arterial bolus input) to the initial (or maximal) value of \(R(t)\):

\[
\text{MTT} = \frac{\int_0^\infty R(t)\,dt}{\max[R(t)]}
\]

Hence, the CBF can be calculated as

\[
\text{CBF} = \frac{\text{CBV} \cdot \max[R(t)]}{\int_0^\infty R(t)\,dt}
\]

It should be noted that an instantaneous arterial bolus input is typically not achievable in practical tracer experiments. Normally, an approximation to the true arterial input function (AIF) is measured and \(R(t)\) is retrieved by deconvolution.

**Related Articles:** Perfusion imaging, Dynamic susceptibility contrast MRI, Arterial spin labelling, Arterial input function (AIF), Cerebral blood volume (CBV), Mean transit time (MTT)


**Cerebral blood volume (CBV)** *(Magnetic Resonance)* The cerebral blood volume (CBV) is the volume of circulating blood in the brain tissue. The blood volume is the sum of the plasma volume and erythrocyte volume. The regional CBV (sometimes denoted rCBV) is often given in units of mL/100 g, i.e. millilitres of blood per 100 g of tissue. According to
Leenders et al. (1990), the regional CBV in normal grey matter is on the order of 5–6 mL/100 g while the regional CBV in normal white matter is on the order of 2–3 mL/100 g.

The determination of the rCBV is based on the principles of tracer kinetics for non-diffusible tracers (Meier and Zierler, 1954; Zierler, 1962) and relies on the assumption that in the presence of an intact blood brain barrier (BBB), the contrast material remains intravascular. When an intravascular plasma tracer is employed, as in dynamic susceptibility contrast MRI, regional CBV can be derived from the time integral of the tissue concentration of contrast agent (C) in combination with the time integral of the corresponding arterial concentration of contrast agent (CArt), i.e. the arterial input function (AIF):

\[
\text{CBV} = \frac{\int_0^\infty (1 - H_{\text{large}}) C(t)dt}{\rho (1 - H_{\text{small}}) \int_0^\infty \text{CArt}(t)dt} = \frac{\int_0^\infty (1 - H_{\text{large}}) C(t)dt}{\rho (1 - H_{\text{small}}) \int_0^\infty \text{AIF}(t)dt}
\]

In order to obtain whole-blood volume in units of mL/100 g, the correction factor \(1 - H_{\text{large}}/\rho (1 - H_{\text{small}})\) is introduced, where \(H_{\text{large}}\) and \(H_{\text{small}}\) are the haematocrit values in large and small vessels, respectively, and \(\rho\) is the brain density.

Finally, the central volume theorem states that the CBV is related to the cerebral blood flow (CBF) and the mean transit time (MTT) according to the relationship CBV = CBF-MTT.

**Related Articles:** Arterial input function (AIF), Dynamic susceptibility contrast MRI, Cerebral blood flow (CBF), Mean transit time (MTT)


Čerenkov effect

(Radiation Protection) Charged particle radiation (beta particles, alpha particles) has kinetic energy given by the classical equation:

\[
E = \frac{1}{2}mv^2
\]

Although relativity states that the velocity of such particles given by this equation can never be greater than the speed of light in a vacuum (the constant \(c\)), the kinetic energy of such particles emitted due to radioactive decay may imply that their velocity is faster than the speed of light for the medium through which the particles are travelling (related to \(c\) by the refractive index of the material). The particles rapidly lose kinetic energy through interactions with electric fields around the atoms of the material, until their velocity is reduced to sub-light speeds for that material.

The energy lost in this interaction process is in the form of UV and visible light, mainly at the blue end of the spectrum, and is called Čerenkov radiation, after the Russian scientist Pavel Čerenkov who first characterised it. It is most often seen in the cooling tanks in nuclear reactor plants and food irradiation facilities.

Čerenkov radiation

(Radiation Protection) The energy lost when charged particles (normally electrons) are slowed down within a transparent absorbing medium is sometimes given off in the form of UV and visible light, mainly at the blue end of the spectrum. This emitted light is called Čerenkov radiation (after the Russian scientist Pavel Čerenkov who first characterised it). It is most often seen in the cooling tanks in nuclear reactor plants and food irradiation facilities.

**Related Article:** Čerenkov effect

Cerrobend®

(Radiotherapy) Frequently, part of the radiation field has to be shielded to avoid irradiating underlying sensitive structures. When individual blocks are designed, low-melting-point alloys are used. Popular products currently in use are Cerrobend® and Ostat alloy®, which are alloys of lead, cadmium, bismuth and tin. They have a melting point of 69°C and 70°C, respectively, although they are usually worked at temperatures of 90°C and 95°C, respectively, since they are easier to pour and mould at these temperatures. Cerrobend has a density of 9.4 g/cm³ at 20°C (~85% of lead density); it consists of 50% bismuth, 26.7% lead, 13.3% tin and 10% cadmium. It is important that the Cerrobend is poured slowly to prevent formation of air bubbles.

Newer materials such as MCP 96® (melting point 96°C, alloy of lead, bismuth and tin) do not contain cadmium, which has relatively high toxicity.

**Related Articles:** Low-melting-point alloy, Block design, Custom blocking


Certification

(General) Certification (professional certification or qualification) is a designation earned by a professional to assure his qualification to perform specific job. Usually, certifications are issued by specific professional body/panel, which assesses the candidate. The main role of such a body is to safeguard the public interest. For example, the UK Institute of Physics and Engineering in Medicine (IPEM) is licensed by the Engineering Council to register Chartered Engineers and by the Science Council to register Chartered Scientists. Usually, a certified professional is listed in a Professional Register and can use post-nominal letters (e.g. CSci - Chartered Scientist, used by many Medical Physicists in the UK).

CFI (colour flow imagine)

(Ultrasound) See Colour flow imaging (CFI)

Chamber response

(Radiation Protection) Chamber response refers to the sensitivity of an ionisation chamber (whether it be operating as an ionisation chamber, proportional counter or Geiger–Müller counter) in detecting the incident radiation, normally measured in terms of the size of the output pulse amplitude from the chamber. A chamber being able to respond to – i.e. to detect lower radiation doses/dose rates will give a larger pulse amplitude and is said to be more sensitive.

Chamber response may also refer to the way the displayed reading on an electrometer of the detection of radiation by an ionisation chamber is affected by a number of parameters. These include radiation-related factors such as the type, energy and fluence rate of the radiation that affect the total charge liberated within the chamber. Other factors affecting the response include chamber-related factors such as the shape, geometry and angular response of the chamber affecting the collection of the liberated charge by the electrodes.
and processed by the electrometer, and environmental factors such as the ambient air temperature and pressure.

The ideal ionisation chamber will have a consistent (or ‘flat’) response to the incident radiation irrespective of changes in any of these parameters. However, in reality, ionisation chambers do not have a flat response and will need to be characterised to compensate for variations due to changes in each parameter. Such characterisation may be done by the manufacturer or as part of the calibration process, comparing the response of the chamber, for example to well-defined radiation quantities and energies at precisely measured ambient temperatures and pressures.

Related Articles: Ionisation chamber, Proportional counter, Geiger–Müller (GM) counters


Characteristic curve
(Diagnostic Radiology) A characteristic curve for radiography film is the graph representing the relationship between optical density and exposure. It is also known as a $H$ and $D$ curve named after the Swiss chemist Ferdinand Hurter (1844–1898) and English chemist Vero C. Driffield (1848–1915), who developed this concept after the Swiss chemist Ferdinand Hurter (1844–1898) and English chemist Vero C. Driffield (1848–1915), who developed this concept for photographic emulsions (Figure C.20).

A specific feature of the characteristic curve is the logarithmic scale used for exposure, usually showing the relative exposure.

The slope of the curve at each point represents the contrast transfer characteristics of the film. A limitation of film is that its scale used for exposure, usually showing the relative exposure.

The toe of the curve at the very low end of the exposure scale represents density not produced by intended exposure. It is the density of the film base plus density, referred to as fog, which may come from a variety of sources such as unwanted exposure from the environment, long age of unprocessed film and problems with film development (chemical fog).

The straighter portion of the curve with a steep slope is the region that produces image contrast. The slope is determined by a combination of film design characteristics and processing conditions. The film gamma (the maximum slope) and average gradient are two parameters used to specify the overall slope of the curve. The film contrast factor, defined as the difference in density for a doubling of exposure, is used to specify the slope at all points along the curve.

Characteristic function
(Diagnostic Radiology) See Characteristic curve

Related Article: K-edge metal filter

Characteristic radiation
(Diagnostic Radiology) Characteristic radiation is the type of x-radiation produced by electron transitions between shells in an atom. The process begins when bombarding electrons enter the x-ray target material or anode and expel electrons from the inner shells with the higher binding energies. These electron vacancies are then filled from the shells with lower binding energies. Because the electrons are moving to a lower energy state, the difference in energy is expelled as an x-ray photon (Figure C.21). The energy of the photon is determined by the specific energy difference between the electron shells, which is a ‘characteristic’ of the specific material. The typical characteristic x-ray spectrum consists of just a few specific photon energies.

Characteristic radiation is a significant proportion of the mammographic beam spectrum. Figure C.22 shows a sketch of a typical mammographic x-ray spectrum, produced by molybdenum anode (note the two lines of its characteristic radiation). Figure C.22 also shows an additional K-edge metal filter (also molybdenum), which absorbs most of the x-rays with energy above 20keV.

Related Article: K-edge metal filter

Characteristic x-ray
(Diagnostic Radiology) See Characteristic radiation

Charge
(General) Electric charge is a basic property of matter carried by some elementary particles and occurs in discrete natural units. This
Charge deposition effect

Charge-coupled device (CCD)

Charge is either positive (+ve) or negative (−ve) in nature and cannot be created or destroyed. Objects that have an excess of one type of charge exert a force on one another when in proximity. Objects that are oppositely charged attract each other, and those with the same charge repel each other. The unit of charge in the SI system is the coulomb (units: m kg s) and the electron has a negative charge of $1.60217733 \times 10^{-19}$ C.


Related Articles: Coulomb, Force electrostatic

Charge deposition effect

(Radiotherapy) If a beam of ionising radiation passes through a material, then charged ions can be produced. These can travel within the material but it may take some time for recombination to take place and this might result in a build up of static charge. Therefore, a residual charge may be present in the material for some time after the irradiation has ceased, which could influence subsequent dose measurements. This effect has to be taken into account in dosimetry, and for this reason, it is recommended that electron dosimetry is carried out using slabs of tissue equivalent material instead of larger blocks since it is easier to remove residual charge from slabs.

Related Articles: Charge, Electrical charge, Electron beam dosimetry

Charge measurement mode

(Radiation Protection) The charge measurement mode is applied in the read-out of a semiconductor detector. The signal (charge) from the detector passes through a charge-sensitive preamplifier and then through a linear amplifier, integral or differential discriminator (single-channel analyser) to counter or count-rate meter.

Related Articles: Germanium detector, Pulse-height analysers (PHAs) for radiation detectors


Charge-coupled device (CCD)

(Diagnostic Radiology) A charge-coupled device is an analogue shift register that is used in many imaging devices. The device was proposed by Eugene Lally at the Jet Propulsion Laboratory in 1961. The first commercial devices were made in 1974. The devices are made so that light falling on the devices deposits charge in an array of pixels. The pixels are then read out in one of several fashions to produce a digital image.

In this process, the data are passed from row to row in the matrix and then is read out – illustrated in the following four images (Figure C.23).

CCDs have replaced analogue traditional video camera tubes in almost all radiology applications and are main sensors in digital photography devices (some modifications being also used in medical imaging). The CCD devices are better, cheaper and less likely to need maintenance (Figure C.24).

Abbreviation: CCD = Charge-coupled device.

Hyperlink: en.wikipedia.org/wiki/Charge-coupled_device
Charge-sensitive preamplifier

(General) A charge-sensitive preamplifier (or amplifier) has an output that is proportional to the electric charge applied to its input.

In radiation sensors, the amount of radiation falling on a sensor in a given time produces a proportional build up of electric charge in the sensor. For this to be read out and amplified, the charge can be sensed in two ways:

1. The charge \( Q \) collects within the sensor and generates a proportional voltage \( V_c \) due to its internal capacitance \( C \), where \( V_c = Q/C \). This signal voltage must then be amplified without draining away the charge – requiring an amplifier with both low drift and with exceptionally high input impedance. Such amplifiers are sometimes known as ‘Electrometer amplifiers’.
2. The charge \( Q \) is allowed to flow into the amplifier as it is generated, where the flow of charge (a current) may be integrated on a separate capacitor and the resultant voltage amplified and buffered.

Whichever technique is used, the amplifier must be low-drift, and some additional circuitry is needed to reset the amplifier to a baseline value by effectively discharging the capacitance holding the charge.

Charged particles

(Radiation Protection) A charged particle is, in physics, either a subatomic particle with an electric charge or an ion. A subatomic particle is one smaller than an atom and there are two broad categories – those that are composite, such as protons, which themselves consist of smaller particles (quarks), and fundamental particles, which have no substructure.

In medical physics, the most common charged particles are electrons (beta particles or negatrons), positrons, protons, alpha particles. Neutrons and neutrinos are subatomic particles but have no charge.

Related Articles: Alpha particles, Beta particles

Charged-particle disequilibrium

(Radiotherapy) If a volume \( V \) containing a medium, homogeneous in density and atomic composition, is uniformly irradiated by photons in any small volume inside \( V \), the number, the energy and the direction of secondary electrons entering and leaving the small volume will be the same if the photon attenuation is negligible. Photons first transfer energy to electrons, which in turn deposit the energy in the medium and the energy carried in and out of the small volume by secondary electrons is equal. This condition inside the small volume is indicated as charged-particle equilibrium (CPE). In practice, the CPE is not possible at any point since the photon fluence is changing through photon interactions.

In regions where the photon fluence varies, causing a difference between the total secondary charged particle energy entering and exiting the volume, CPE does not exist. Cases of charged-particle disequilibrium are:

1. Regions in the vicinity of a point radiation source where the photon fluence is changing rapidly with distance.
2. Regions in close proximity to boundaries between media of different composition and where the photon energy is high enough that the range of secondary electrons is no longer negligible compared to the mean free path of the photons.
3. Regions where there is the presence of inhomogeneous electric or magnetic field.

An evident situation of charged particle disequilibrium is in the build-up region of a beam where the dose increases before decreasing exponentially. The uniformity of the photon fluence is also not satisfied near the edges of a finite beam at distances between the beam edge and the point under consideration larger than the maximum secondary electron range. This particular disequilibrium is called lateral disequilibrium and it is obtained at the central axis of the beam when the lateral electron range exceeds the beam radius. This effect is particularly evident in small field size such as in stereotactic radiosurgery.

A transient charged-particle equilibrium (TCPE) exists at all points within a volume in which the dose \( D \) is proportional to collision kerma \( K_{\text{coll}} \), while for CPE \( D = K_{\text{coll}} \). This situation is obtained on the axis of a broad high-energy photon beam where the photon fluence is not uniform because of the geometric divergence and the photon attenuation.

Abbreviations: CPE = Charged-particle equilibrium and TCPE = Transient charged-particle equilibrium.

Related Articles: Charged-particle equilibrium, Collision kerma, Build-up region, Electron maximum range, Electron practical range, Percentage depth dose

Charged-particle equilibrium

(Radiotherapy) Charged-particle equilibrium (CPE) is established in a volume when the energy carried into the volume by the charged particles is balanced by the energy carried out of the volume by the charged particles.

If a volume \( V \) containing a medium, homogeneous in density and atomic composition, is uniformly irradiated by photons in any small volume inside \( V \), the number, the energy and the direction of secondary electrons entering and leaving the small volume will be the same if the photon attenuation is negligible. Photons first transfer energy to electrons, which in turn deposit the energy in the medium and the energy carried in and out of the small volume by secondary electrons is equal. This condition inside the small volume is indicated as CPE.

When CPE is established, the collision kerma \( K_e \) is equal to the dose \( D \):

\[
D = K_e \quad \text{(CPE)}
\]

Charged-particle equilibrium is not established for thin layers of materials irradiated by photons as the maximum range of the secondary electrons produced is larger than the material dimensions so that part of the electron energy is lost simply by the electrons leaking out of the material.
Charged-particle equilibrium is also not established in a bulky block of high atomic number Z material, with a gamma ray source distributed all through the volume of the block. The block dimensions are large with respect to the range of the Compton electrons produced by the gamma rays but small relative to the attenuation length of the gamma rays. The atomic number of the material is sufficiently high so that an appreciable amount of energy of the electrons is transformed into Bremsstrahlung radiation, which is then lost to the material and escapes from the block.

**Abbreviation:** CPE = Charged-particle equilibrium.

**Related Articles:** Charged-particle disequilibrium, Collision kerma, Build-up region, Electron maximum range, Electron practical range, Percentage depth dose

### Charged-particle therapy

**Radiotherapy** The term charged-particle therapy is generally used to mean therapy with charged hadrons – specifically protons or light ions (rather than using electrons, which are not hadrons).

**Related Articles:** Hadron therapy, Heavy particle beams, Ion therapy, Proton therapy

### Check source

**Radiotherapy** A radioactive check source is recommended for operational and constancy checks of the dosimetry systems (ionisation chamber and measuring electrometer) used for the calibration of the radiation beams. The check source permits an exposure of the dosimetry system in a fixed and reproducible geometry. Generally, the test device contains a beta-emitting $^{90}$Sr/$^{90}$Y radioactive source because of the radioisotope long half-life. The typical initial activity of the source is approximately 30 MBq housed in a shielded container. The check sources are available for cylindrical and parallel-plane ionisation chambers (Figure C.26).

The ionisation chamber and electrometer performances are checked by using a stopwatch to determine equal exposure in the isotope device. The average reading of the dosimetry system is corrected for temperature and barometric pressure and then compared with an expected value that is the reading obtained on occasion of the calibration of the dosimetry system corrected for the radioactive decay of the check source using the 28.78 year half-life of $^{90}$Sr. If the measured exposure value is within $\pm0.5\%$ of the expected or standard value, the ionisation chamber and electrometer combination is considered reliable (Figure C.27).

An alternative method for operational and constancy checks of the dosimetry systems is to use $^{60}$Co equipment, when available, performing the exposure measurements, in air or in a phantom, at an accurately reproducible distance from the source in a fixed irradiation geometry.

### Chemical dosimetry

**Radiotherapy** The objective of chemical dosimetry is to determine the absorbed dose in an irradiated material from the chemical changes produced by the radiation. Any system, in which a well-characterised and measurable change takes place in a chemical property, may in principle be used as a chemical dosimeter. However, the chemical system, as with any other type of dosimeter, must meet a number of requirements and therefore, even if a great number of chemical systems have been studied, it is possible to characterise a single universal chemical system. Solid, liquid and gaseous systems have been introduced and their application depends on the specific studied problem. Examples of solid dosimeter are plastic film dosimeters, systems...
incorporating organic dyes, alanine dosimeters and photographic films. Most liquid dosimeters are based on dilute aqueous solutions of various compounds. After the irradiation, the aqueous chemical dosimeters can be analysed by titration or light absorption. One of the most studied liquid chemical systems is the Fricke dosimeter in which ferrous ions in a sulphate solution are oxidised by the action of radiation. Other liquid dosimeters are based on ceric sulphate, oxalic acid or on a combination of ferrous sulphate and cupric sulphate. A recent advancement of chemical dosimetry is the development of gel dosimetric systems in which the nuclear magnetic resonance (NMR) relaxation properties of irradiated gels infused with conventional Fricke dosimetry solutions are used to determine the Fricke gel absorbed dose. Fe$^{2+}$ ions dispersed throughout the gel matrix are converted by radiation to Fe$^{3+}$ ions, with a corresponding change in the paramagnetic properties that may be quantified using NMR relaxation measurements or optical measurement techniques. Because of the limitations of those dosimeters related with the ion diffusion in the gel, alternative polymer gel dosimeters have been proposed. Polymer gel monomers, which are usually dispersed in an aqueous gel matrix, undergo a polymerisation reaction as a function of absorbed dose. The formation of polymers induced by the radiation influences the NMR relaxation properties and results in other physical changes that may be used to quantify the absorbed dose. As the polymerisation is inhibited by oxygen, all free oxygen has to be removed from these gels and this makes their use troublesome. Recently, a new polymer gel dosimeter formulation has been introduced in which oxygen is bound in a metallo-organic complex, thus removing the problem of the oxygen inhibition. Quantitative techniques for measuring dose distributions in these dosimeters include also optical and x-ray computer tomographies, vibrational spectroscopy and ultrasound.

*Abbreviation:* NMR = Nuclear magnetic resonance.

**Chemical exchange**

(Radiation Protection) Chemical exchange with ionic resins (ion-exchange method) is used in nuclear medicine for the removal of some elements from the mixture (e.g. radioactive waste). A sample of radioactive solution passes through ion-exchange resin. As a result the purified solution is separated from the radioactive waste.

(Figure C.28)

There are two kinds of ion-exchange resins:

- **Cation-exchange:** R-H + Na$^+$ → R-Na + H$^+$
- **Anion-exchange:** R-OH + Cl$^-$ → R-Cl + OH$^-$

The chemical exchange with ionic resins (ion-exchange method) is very useful in radiochemistry, nuclear medicine and in the removal of some elements from the mixture.

**Related Articles:** Radioactive materials, Radioactive sources, Radioactive waste


**Chemical purity**

(Nuclear Medicine) Chemical purity is the amount of undesirable chemical constituents in a radiopharmaceutical. This measurement is important to ensure that the presence of any chemical material with potential physiologic, pharmacologic or toxic effects is at or below an acceptable limit. The testing should be performed on any chemical substance that is used or formed during the synthesis or production of radiopharmaceuticals, for example chemical impurities, unlabelled components, reagents and by-products.

An example of a chemical impurity significant for nuclear medicine pharmacy is the aluminium ion, Al$^{3+}$, that may be present in the eluate from a $^{99m}$Tc generator. Measurement of Al$^{3+}$ concentration is usually performed by the use of a colourimetric spot test and may be carried out regularly.

Another example is trace metals such as Fe in $^{111}$InCl solutions, which can significantly reduce the labelling efficiency of $^{111}$In – biomolecules and blood cells.

A third example is carrier iodine in sodium–radioiodine solutions, which can compete with the radioactive iodine during iodination processes, resulting in poor labelling efficiency or interference with the uptake in the thyroid of tracer amounts of radioiodide.

**Related Articles:** GMP, Quality control, Radionuclide purity, Radiochemical purity, Biological purity


**Chemical quenching**

(Radiation Protection) Quenching in liquid scintillation detectors causes the reduction of counting efficiency as a result of diminishing the amplitude and number of scintillations. There are several types of quenching: chemical, colour, dilution and optical. The chemical quenching is caused by chemical composition of the radioactive sample added to the liquid scintillation solution for counting. The chemical composition of the sample can change either the energy transfer from the detected radiation to the scintillating solution or the transfer of light from the scintillating solution to the photomultiplier (PM) tube.

Quenching should be taken into account to obtain accurate counting of samples. There are three methods used for this purpose:

1. **Internal standard method** – sample is measured first, then a known amount of the same radioisotope is added and the sample is measured again.
2. **Channels ratio method** – two pulse-height analysers (PHA) are applied to describe the shape of the quench curve. The ratio of the counts (Figure C.29)

$$R = \frac{C_1}{C_2}$$

will decrease with an increase in quenching.

---

**Abbreviation:** Cation-exchange: R-H + Na$^+$

**Abbreviation:** Anion-exchange: R-OH + Cl$^-$

**Figure C.28** Example of a chemical exchange with use of an ion-exchange resin (membrane).
The effective field at the nucleus is generally less than the applied field by a fraction $\sigma$; thus, the effective field $B_{\text{eff}}$ at the nucleus is

$$B_{\text{eff}} = B_0 (1 - \sigma)$$

$\sigma$ is related to the Larmor frequency $\omega_0$ as follows:

$$\omega_0 = \frac{\gamma}{2\pi} B_0 (1 - \sigma)$$

and to chemical shift as follows:

$$\delta = 10^6 (\sigma_{\text{ref}} - \sigma_{\text{sample}})$$

The frequency at which a given nucleus comes to resonance depends therefore not only on the strength of the applied field and the gyromagnetic ratio of the nucleus but also on the molecular environment of the nucleus. Therefore, for a given nucleus, it is observed that the atoms in different chemical environments give MR signals at different characteristic value of the applied field.

**Related Article:** MRS

### Chemical shift artefact

*Magnetic Resonance* Chemical shift is caused by the inherent differences in the chemical environment of protons in water molecules and those in fat molecules. When subject to a magnetic field, protons in a fat molecule will have a lower resonant frequency than those in water molecules due to its molecular environment. MR systems are most frequently tuned to the resonant frequency of water. When frequency encoding is applied to tissue, it causes the fat signal to be shifted a number of voxels from its actual position in the frequency-encoding direction. The misregistered fat signal appears in the form of light or dark bands across the image.

The amount of chemical shift artefact experienced for a particular image depends mainly on the receive bandwidth used. There is a balance between increasing bandwidth to minimise chemical shift effects and reducing signal to noise ratio as more noise is included with the desired signal.

The chemical shift of the magnet depends on the strength of the static magnetic field, $B_0$. The chemical shift between water and fat for a 1.5 T magnet is 3.5 ppm. This typically translates to a shift on the order of a few voxels.

Another type of chemical artefact, called the phase cancellation artefact, exists in voxels containing both fat and water. This appears as a black outline on the image between the fat and water boundaries. It is caused by the fat and water dephasing after the initial excitation due to their difference in resonant frequencies. To reduce this artefact, in-phase spin-echo pulse sequences are used.

**Abbreviation:** PPM = Parts per million

**Related Article:** Frequency encoding

### Chemical shift imaging (CSI)

*Magnetic Resonance* Chemical shift imaging (CSI), otherwise known as magnetic resonance spectroscopic imaging (MRSI), combines spatial encoding techniques drawn from MRI with magnetic resonance spectroscopy (MRS) in order to produce one-, two- or three-dimensional arrays of spectra.

The advantage of CSI lies in its ability to collect data from multiple voxels simultaneously. Thus, it is not necessary to choose the optimal volume of interest (VOI) location prior to signal acquisition, and it is possible to map variations of metabolite signals within organs and between diseased and normal-appearing regions.
Conventional CSI uses similar pulse sequences to those employed in MRI, except that frequency encoding is avoided because imposition of a gradient during signal acquisition would preclude extraction of spectroscopic information using Fourier analysis. Therefore, two-dimensional (2D) CSI uses slice selection followed by phase encoding on two orthogonal axes, and three-dimensional CSI requires phase encoding on all three axes. These requirements make CSI a slow and/or low-resolution technique, although adoption of fast imaging techniques developed in the context of MRI has done much to redress this in recent years (Figure C.30).

CSI is a technique with a long history, dating back to the 1970s before the development of 2D Fourier MRI. However, hardware and pulse sequence limitations precluded widespread implementation on clinical MRI systems until the 1990s. Not least among the requirements, given the prevalence of proton (1H) spectroscopy, is excellent static field uniformity, allowing water suppression over a large volume. This necessitates shimming prior to each examination and effectively limits applications of the technique to a few regions of the body, such as the brain and prostate. This problem is less serious for CSI using nuclei other than hydrogen. Because time constraints limit the number of phase-encoding steps that can be performed in CSI, data sampling is sparse and the point spread function correspondingly poor. This leads to ‘Fourier bleed’ of signal between voxels. A particularly serious manifestation of this in brain CSI is bleed of intense lipid signals from the margin of the skull into voxels within the brain, where they can obscure metabolite signals of interest. For this reason, CSI is often combined with spatial presaturation bands angled so as to eliminate this lipid signal.

CSI data can be displayed in the form of arrays of spectra, but often, it is often helpful to extract the signal intensity of each resonance from each spectrum and display these signal intensities as a metabolite map or in the form of metabolite signal ratios, often using colour overlay on a structural MR image. However, while such images are useful qualitatively, it is better to regard the spectra as the base information for quantification since derived parametric maps may be subject to a range of errors (Figures C.31 and C.32).

Related Articles: Chemical shift, Frequency encoding, Magnetic resonance imaging, Magnetic resonance spectroscopy, Phase encoding, Single voxel spectroscopy


Chemotaxis targeting

(Nuclear Medicine) Chemotaxis refers to the process in which bodily cells, bacteria or multicellular organisms direct their movement according to certain chemicals in their environment. The chemotaxis process is used to study inflammatory processes in which radiolabelled leukocytes are directed to the inflammatory area by the chemical products produced there.
Chest radiography

(Diagnostic Radiology) Chest radiography is one of the most frequently performed x-ray examinations. Chest radiography presents a wide dynamic range from bones to air-containing lungs. The patient-to-patient variability (anatomic noise) in this region is also significant. The high dynamic range of contemporary digital x-ray systems present far better results than screen-film radiography, but tomographic methods as CT and MRI are far more effective in detection of low-density lung lesions lying under high-density bones (ribs cover $\sim 2/3$ of the lung). Cardiac examinations in the region are made with angiographic equipment.

Physical Contrast in the Chest: The lungs that are partially filled with air form a low-density background that appears dark in the usual radiographic display. Most other structures (ribs, heart, abnormal masses within the lungs, etc.) that are more dense than the lungs appear as bright areas. The chest radiograph is a display of the variations in physical density that are within the region (Figure C.33).

General Imaging Techniques for Chest: Conventional chest radiography is performed with an x-ray spectrum containing high-energy photons to enhance penetration through the body. A potential problem in chest radiography is that the very high physical contrast within the chest, described earlier, can produce a wide range of exposure values to the image receptor that extend beyond the receptor’s range of good contrast transfer. This must be addressed both through the optimisation of the x-ray spectrum and the selection of receptor and display contrast characteristics.

An x-ray spectrum that is generally optimum for chest imaging is produced with a relatively high KV value and more beam filtration than used for most other radiographic procedures. This reduces the high contrast between the low- and high-density areas and also produces good penetration and visibility through the ribs (Figure C.34).

For film chest radiography, wide latitude (or dynamic range) film is generally used. This reduces the potential of the wide density range in the body producing both areas of over- and underexposure within the image.

Digital radiography with wide dynamic range receptors and the capability to digitally process the images and apply windowing reduces the problems produced by the high physical contrast within the chest. A number of new investigations of chest imaging point out that compared with high KV classical film radiography, digital radiography produces better images with lower KV (on the order of 80 vs. 120 KV).

Related Article: Abdominal imaging


Hyperlink: Sprawls foundation: www.sprawls.org/resources

Chi-square test

(General) The Chi-square $\chi^2$ test is a statistical test useful to determine the presence of a random error other than the expected variation of a Poisson distribution in a set of measurements. Examples of measurements where this test is useful are in the determination of faulty instrumentations and the normal variation between patients, animals, etc. The first step of the test is to obtain a series of measurements, which should be at least 20. The second step is to compute the mean value of the measurements

\[ \bar{N} = \frac{1}{n} \sum_{i=1}^{n} N_i \]

and the quantity

\[ \chi^2 = \sum_{i=1}^{n} \frac{(N_i - \bar{N})^2}{\bar{N}} = \frac{(n-1)SD^2}{\bar{N}} \]
The $\chi^2$ square value can be referred to a graph provided in most statistics text books. Locate the $P$-value closest to the position with coordinates $n$ and $\chi^2$. The $P$-value is the probability that random variation from a Poisson distribution would equal or exceed the $\chi^2$-value. A $P$-value close to 0.5 suggests that the calculated $\chi^2$-value is in the middle range, expected for a Poisson distribution.

A lower $P$-value indicates that there is a high probability that there is an additional source of error present. A higher $P$-value indicates that there is less variation than expected from a Poisson distribution but should also be a cause for concern. Typically, a $P$-value between 0.05 and 0.95 is acceptable. $P$-values above 0.99 and below 0.01 indicate that there is likely to be a fault in the measurement system and $P$-value above 0.95 and below 0.05 indicates that the measurements should be considered inconclusive and repeated.

$\chi^2$ is in itself a statistically variable quantity and should therefore be treated with caution. The standard deviation for $n \sim 20$ is about 25% and for $n \sim 100$, it is 15%. If values are close to the critical $\chi^2$ values, the experiment is recommended to be repeated.


**Choline**

(*Magnetic Resonance*) Choline (Cho) is a chemical compound that is detected in in vivo proton ($^1$H) NMR spectra of a number of organs (Figure C.35).

The 3.22 ppm trimethylamine resonance frequently, but somewhat loosely, attributed to Cho in a proton spectrum in fact arises from a number of related choline-containing compounds: phosphocholine, glycerophosphocholine and betaine, as well as Cho itself. The peak is therefore more properly ascribed to choline-containing compounds, or total choline (tCho).

Choline-containing compounds play a role in membrane synthesis, and consequently, the tCho peak is elevated in malignant tumours due to rapid cell division (Figure C.36).

**Chromium-51 [51Cr]**

(*Nuclear Medicine*)

<table>
<thead>
<tr>
<th>Element: Chromium</th>
<th>Isotopes: 18 &lt; $N$ &lt; 38</th>
</tr>
</thead>
<tbody>
<tr>
<td>Atomic number ($Z$): 24</td>
<td>Mass number ($N$): 27</td>
</tr>
<tr>
<td>Symbol: $^{51}$Cr</td>
<td>Production: Usually reactor $^{51}$Cr($n$, $\gamma$)$^{51}$Cr</td>
</tr>
<tr>
<td>Daughter: $^{51}$V</td>
<td>Half-life: 27.7 days</td>
</tr>
<tr>
<td>Decay mode: $\beta^-$ decay</td>
<td>Radiation: Gamma, conversion electrons, Auger electrons, characteristic x-ray photons</td>
</tr>
<tr>
<td>Gamma energy: 320 keV (10%)</td>
<td>Dose rate from 1 MBq: 0.064 mSv/h at 30 cm (point source); 0.0054 mSv/h at 1 m (10 mL vial)</td>
</tr>
<tr>
<td>Absorption (HVL): 2 mm lead (note the very low photon yield)</td>
<td>Biological half-life: 8 h (30%), 10–160 days</td>
</tr>
<tr>
<td>Critical organ: Liver, plasmaleptins</td>
<td>$\text{ALI}_{\text{max}}$ (50 mSv): 700 MBq</td>
</tr>
<tr>
<td>Absorbed dose: 0.6 mGy/MBq liver</td>
<td>Effective dose: 0.038 mSv/MBq (oral); 0.021 mSv/MBq (inhalation)</td>
</tr>
</tbody>
</table>

**Clinical Applications:** $^{51}$Cr is labelled to the chelate EDTA useful for glomerular filtration rate measurements (blood samples) in vitro. $^{51}$Cr-labelled RBCs have commonly been used for measurements of red cell mass, survival studies of RBC and platelets and plasma volume.


**CIF (contrast improvement factor)**

(*Diagnostic Radiology*) See Contrast improvement factor (CIF)

**Cine film**

(*Diagnostic Radiology*) Cine film is mainly used in cineangiography to record angiographic x-ray examination with cine camera (taking the image from the output of the image intensifier through special optics). This roll film is mainly with two widths: 35 mm (more often) and 16 mm. The wider film covers a larger area and is mainly associated with greater dose. The film is processed by special processing machines and is further viewed by projector (as movie film).
Contemporary angiographic systems do not use cine film anymore, but directly record the dynamic digital image in the system memory.

**Related Articles:** Cineangiography, Cineradiography, Cinefluoroscopy, Angiography

**Cine loop**

*(Diagnostic Radiology)* Cine loop is a method of observation of cine films, where a section of the film is repeatedly observed. The method is common in x-ray angiographic cardio examinations, where movement of the cardiac muscle has to be examined at a specific period of the cardiac cycle. The method is also used (often with the same name) in digital radiography, where a sequence of digital frames is looped and after the end of the last selected image, the first image is again addressed and the whole sequence repeats.

**Related Articles:** Fluoroscopy, Cinefluoroscopy, Cineangiography, Digital subtraction angiography

**Cine MRI**

*(Magnetic Resonance)* Cine MRI techniques are designed to acquire a series of individual MRI image frames and display them sequentially as a movie. The primary application of cine MRI is in cardiac imaging where a cine acquisition is used to display a movie of the heart beating over a single cardiac cycle (see Figure C.37).

The basic building block of any movie is a frame. The duration of a frame determines the temporal resolution of the movie. Typical frame durations useful in cardiac cine MRI are on the order of 50 ms.

In a real-time cine acquisition, each acquired frame corresponds to a single snapshot of the heart at that instant. Real-time acquisitions are technically demanding as k-space for each frame must be completely filled in ~50 ms, while still achieving adequate resolution and SNR. Through technical developments such as the application of parallel imaging techniques, real-time cardiac for routine clinical cardiac applications is likely to become feasible. To date however, most cine MRI acquired in clinical practice is not real time. While the final movie appears to show the time course of a single beat, the actual data acquired to generate each frame of the movie are collected over many heartbeats.

![Cine MRI images of short axis of heart.](image1)

**Figure C.37** Cine MRI images of short axis of heart.

This type of acquisition is illustrated in Figure C.38. Each movie frame has a corresponding k-space that must be filled. In a real-time acquisition, each k-space frame would be filled completely in a single frame duration and the entire movie collected in a single heartbeat. More generally, only a few lines in k-space will be filled for each frame in every heartbeat. The number of lines filled per heartbeat per frame is called the views per segment. Segmented k-space acquisitions allow total scan times that are within the breath-holding ability of most patients— an important consideration in reducing movement artefacts.

An ECG trace is acquired during scanning to provide a temporal reference to the phases of the beating heart. In prospective gating, the ECG directly triggers acquisition to ensure that k-space lines for each movie frame are acquired at the same phase of the cardiac cycle in each heartbeat. In retrospective gating, the MRI data are acquired continually and rebinned appropriately into k-space frames after the scan by reference to the recorded ECG.

The main sequence type used in cardiac cine imaging is balanced FFE (i.e. a true FISP), which provides good inherent blood/myocardium contrast. Prior to the availability of balanced FFE sequences, spoiled gradient-echo sequences (e.g. FLASH-type sequences) were used. Blood/myocardium contrast with spoiled gradient-echo sequences is achieved through the bright blood time-of-flight technique. Balanced techniques have the advantage of faster acquisition times and independence of contrast on blood.
Cinescopic or radiographic mode of operation of the x-ray generator. If the the quick exposure of the film requires higher light intensity, hence higher x-ray beam intensity.

The cine camera uses an electronic synchronous motor to move the cine film. The number of frames per second is based on the power supply of the camera. For example, if 60 Hz is used, the camera can record with speed up to 60 fps, or, respectively, 30 or 15 fps.

Video recording of the fluoroscopic image had been used in the past, but its quality is lower than the cine film quality. Contemporary digital fluoroscopic units (with II or with flat panel detector) record the digital image directly and allow easy image manipulation and record. Their quality is rapidly moving to outperform cine image quality.

**Related Articles:** Fluoroscopy, Biplane cine system, Cineradiography, Cineangiography

### Cineradiography

(Diagnostic Radiology) Cineradiography is the method of recording angiographic x-ray examination. As this system records images of blood vessels filled with contrast media (most often iodine-based), the x-ray spectrum has to comply with the absorption K-edge of iodine. Cineangiography can use either the fluoroscopic or radiographic mode of operation of the x-ray generator. If the radiographic mode is used, the kV of the x-ray pulses can vary above or below the K-edge, thus allowing dual-energy subtraction (hence better contrast resolution).

Cineangiography systems used in cardioangiography use synchronisation of the imaging with the electrocardiography of the patient. This allows recording at different phases of the cardiac cycle. Contemporary systems do not use cine film, but directly record the digital image in the system memory.

**Related Articles:** Fluoroscopy, Biplane cine system, Cineradiography, Cinefluoroscopy, Angiography, Dual-energy imaging

### Cinefluoroscopy

(Diagnostic Radiology) Cinefluoroscopy is a system/method allowing the recording of dynamic images from fluoroscopy to cine film. During cinefluoroscopy, the x-ray generator usually operates in continuous fluoroscopic mode. The cine camera is attached to the output of the image intensifier (II) through special optics. This can be fibre optic type or using lenses (often called tandem optics). The latter uses semi-transparent mirror to split the light output from the II to two beams – one going to the TV monitor, the other one to the cine camera (and often a third output to spot-film radiography) – see Figure C.39.

The cine film used in such a system is 16–35 mm. As these systems are mainly used for cardioangiography, the 35 mm camera and film are preferred as these have better resolution. However, the patient dose in all cine recording x-ray systems is very high, as the quick exposure of the film requires higher light intensity, hence higher x-ray beam intensity.

The cine camera uses an electronic synchronous motor to move the cine film. The number of frames per second is based on the power supply of the camera. For example, if 60 Hz is used, the camera can record with speed up to 60 fps, or, respectively, 30 or 15 fps.

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**Related Articles:** Fluoroscopy, Biplane cine system, Cineradiography, Cineangiography

### Further Reading


### Circuit breaker

(General) Circuit breaker is an electrical switch, which interrupts the electric current in a circuit when the current becomes too high, caused by overload or short circuit. Unlike a fuse, a circuit breaker can be reset (either manually or automatically) to resume normal operation after it has been tripped. When the current is high enough to operate a trigger device in a breaker, a pair of contacts conducting the current are separated by preloaded springs or other mechanism. Generally, a circuit breaker registers the current either by the current’s heating effect or by the magnetism it creates in passing through a small coil. Circuit breakers are made in varying sizes, from small devices that protect an individual household appliance up to large switchgear designed to protect high-voltage circuits feeding an entire city.

### Circuit(s), electrical

(General) An electrical circuit is a term that refers to any interconnected network of electrical components. Current can only flow when a path or circuit exists between the two terminals of any power source.

A circuit diagram or schematic is used to describe the components and how they are interconnected. Solid lines represent the...
interconnection of components using a set of internationally agreed symbols for the commonly occurring devices, and their properties are usually displayed in numbers and units alongside each component (Figure C.40).

**Related Articles:** Resistor, Capacitor, Diode

**Further Reading:** ANSI standard Y32 (also known as IEEE Std 315); IEC 60617 (also known as British Standard BS 3939)

### Circular polarisation

(Magnetic Resonance) See Circularly polarised (CP)

### Circularly polarised (CP)

(Magnetic Resonance) If the $B_1$ field vector (i.e. the magnetic component of the transmitted RF pulse) at a point in space undergoes constant rotation in a plane, then the field is said to be circularly polarised at that point.

A circularly polarised $B_1$ field is physically equivalent to the summation of two orthogonal, linearly polarised fields in phase quadrature (Figure C.41):

$$B = B_1 \sin \omega t \ y + B_1 \cos \omega t \ x$$

**Related Articles:** $B_1$, Linearly polarised, CP coils

### Circularly polarised coil (CP coil)

(Magnetic Resonance) See Quadrature coil

### Classic (coherent) scattering

(Radiation Protection) See Coherent scattering

### Classified workers

(Radiation Protection) The term ‘classified workers’ refers to those workers who are employed in work activities involving normal exposure to ionising radiation. The classification of such workers is based on the evaluation of the individual activity (and relative risk) and measurements of the work situation. Detailed working conditions, responsibilities, duties and monitoring arrangements are clearly specified in the national set of radiation protection standards. The limits only refer to work activities and shall not include medical exposures and background.

**Related Articles:** Dose limits, Occupational exposures

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**Clearance of tracers**

(Nuclear Medicine) This entry refers to the clearance of tracers from blood in the capillaries into tissue. The amount of tracer extracted from the blood is the product of the extraction fraction and the blood flow through the capillaries. The unit is the same as for a flow, i.e. mL/min or mL/min/g. The increase in tracer extraction due to an increase in blood flow offsets the decrease in extraction factor; hence, the net tracer delivery is proportional to blood flow.

In tracer kinetic modelling, clearance is also used to describe the clearance of tracer from tissue to blood.


**Related Articles:** Tracer kinetic modelling, Tracer, Tracer delivery

### Clinical target volume (CTV)

(Radiotherapy) The clinical target volume (CTV) describes the full extent of malignant growth to be treated with radiotherapy. This includes any subclinical microscopic growth that is not visible with standard imaging techniques. It is often determined by adding a margin to the gross tumour volume (GTV), the margin based on anatomy and biological considerations. Functional imaging techniques such as PET may add information on the extent of the margin needed. The CTV may include regional lymph nodes. Margins will be added to the CTV to create the planning target volume (PTV).

The use of CTV as a planning volume was proposed by the ICRU in Report 50 (with addendum 62). This report provides a common framework on prescribing, recording and reporting therapies, with the aim to improve the consistency and inter-site comparability. It details the minimum set of data required to be able to adequately assess treatments without having to return to the original centre for extra information (Figure C.42).

**Abbreviation:** CTV = Clinical target volume.

**Related Articles:** ICRU, Clinical target volume (CTV), Planning target volume (PTV), Treated volume, Irradiated volume

**Further Readings:** ICRU (International Commission on Radiation Units and Measurements). 1993. Prescribing, reporting

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**FIGURE C.40** Common symbols used in electrical circuits.

**FIGURE C.41** Circularly polarised $B_1$ field. A circularly polarised field decomposes into two orthogonal, linearly polarised fields in quadrature (denoted $B_{1x}$ and $B_{1y}$ in the diagram).

**FIGURE C.42** Definition of target volumes as in ICRU 50.

Clock frequency
(General) See Clock pulse

Clock pulse
(General) A clock pulse is an electric signal used to coordinate the actions of two or more digital electronic circuits. The clock pulse can be either in high or low state. The electronic circuits may synchronise at the rising edge, falling edge or both in the rising and in the falling edges of the clock pulse. Clock pulses are continuous, precisely spaced changes in voltage, also known as clock signals. Most often, they are in the form of a square wave with a 50% duty cycle, usually with a fixed, constant frequency.

Clutter
(Ultrasound) Clutter is a term used to describe unwanted echoes from electronic systems. The term is most commonly used not only in radar applications but also to describe noise in B-mode and Doppler imaging.

Clutter in colour flow imaging arises from the high-intensity signals from the vessel wall and surrounding tissue. These are usually relatively slow-moving structures and filters are used to remove the high intensity but low frequency shift echoes. However, these filters will also remove signals from low-velocity blood flow and engineers face challenges in applying and setting filters appropriate to specific applications.

Related Articles: Wall filter, Grating lobes, Side lobes

CMUT
(Ultrasound) See Capacitive micromachined ultrasound transducer.

Coaxial cable
(General) A coaxial cable is a cable consisting of two conductors laid concentrically along the same axis. The inner conductor is surrounded by a tubular flexible dielectric insulator, which is in turn surrounded by fine woven wire, or a thin metallic foil acting as a shield. The whole cable is wrapped in a protective plastic sheathing. The term coaxial comes from the inner conductor and the outer shield sharing the same geometric axis. The current in the inner conductor draws the current in the outer conductor towards the centre rather than letting it dissipate outwards and thus the associated current flow is restricted to the adjacent surfaces of the inner and outer conductors. As a result, coaxial cable has very low radiation losses and low susceptibility to external interference and effectively guides signals with low emission along the length of the cable. Coaxial cables are often used as a transmission line for radio frequency signals. Because they can carry a large number of signals simultaneously, coaxial cables are widely used in cable television systems.

Cobalt
(General) Cobalt (Co) is a transition metal with atomic number 27. It is a ferromagnetic metal and found in compounds in nature. Cobalt has many isotopes ranging from \(^{50}\text{Co}\) to \(^{73}\text{Co}\); \(^{59}\text{Co}\) is stable. The remaining are radioisotopes, of which \(^{59}\text{Co}\) and \(^{60}\text{Co}\) are commonly used in medical physics.

\(^{60}\text{Co}\) in Radiotherapy: Cobalt-60 has a radioactive half-life of 5.27 years. It decays into \(^{60}\text{Ni}\) with the emission of beta electrons with a maximum energy of 320 keV, and two gamma rays with energies of 1.17 and 1.33 MeV. The electrons are absorbed into the source capsule, while the high-energy gamma rays can be used for radiation therapy. The 5.27 year half-life means that cobalt sources need replacing at regular intervals.

\(^{57}\text{Co}\) in Nuclear Medicine – Schilling’s Test: Cobalt-57 has a radioactive half-life of 271.8 days. It decays into \(^{57}\text{Fe}\) by electron capture, which also emits gamma rays at 122.1 keV. It is used in nuclear medicine for the Schilling test, which can help to determine the uptake of vitamin B12.

Related Article: Cobalt unit

Cobalt unit
(Radiotherapy) The \(^{60}\text{Co}\) radionuclide in a cobalt unit decays with a half-life of 5.26 years into \(^{60}\text{Ni}\) with the emission of electrons (beta particles), with a maximum energy of 320 keV, and two gamma rays with energies of 1.17 and 1.33 MeV. The emitted gamma rays constitute the therapy beam, while the electrons are absorbed in the cobalt source or the source capsule, where they produce relatively low energy and essentially negligible Bremsstrahlung x-rays and characteristic x-rays.

Cobalt-60 teletherapy units developed in the 1950s provided the first practical megavoltage treatment units. At that time, this type of treatment offered a significant step towards high-energy treatment and meant that \(^{60}\text{Co}\) teletherapy was at the forefront of radiotherapy for many years. The advantages of cobalt units are that they have a high-energy gamma ray emission, a long half-life and high specific activity and simple means of production. Although linear accelerators have taken over in most centres because of the sophisticated treatment capability, cobalt units are still in use in many places throughout the world because they are far simpler, easier to maintain and cheaper to purchase (Figures C.43 through C.45).

Related Articles: Caesium, Gamma knife

FIGURE C.43 A typical cobalt-60 treatment unit showing the treatment couch and the control pendant suspended from the ceiling.
Coded aperture

Pinhole apertures with several holes arranged according to specific binary arrays are collectively called coded apertures. Coded apertures were originally used in astronomy, but have recently been introduced in nuclear medicine applications, mostly for preclinical imaging. The advantage over conventional pinholes is an increased efficiency due to a large open fraction without a loss in spatial resolution. The signal to noise ratio depends on the source spatial distribution and on the specific coded aperture pattern used. In a coded aperture imaging setting, assuming that the source consists of several point sources, each point source casts a shadow of the pattern on the detector. The size of the projected shadow depends on the geometrical setting (object to aperture, object to detector and aperture to detector distance). Each coded aperture pattern is associated with a decoding mask, and in order to decode the projected overlapped shadows, the acquired image is convolved with a decoding matrix. The dimensions of the decoding mask are also dependent on the geometrical settings. Common coded apertures used in nuclear medicine applications include uniformly redundant arrays (URA) and modified uniformly redundant arrays (MURA). Due to practical fabrication constraints, a row and column representing solid material is introduced between each row/column in the mask. This is called a ‘no two holes touching’ (NTHT) technique.

A third family of patterns, a special case of MURA, has a special property, namely that if rotated 90°, the pattern is the original patterns anti-mask, i.e. each hole is now an open pinhole and vice versa (Figure C.46). A reconstructed image can be obtained by summation of two images, one acquired with the original mask and one with the anti-mask. Due to the nature of the coded aperture arrays, the method is subjected to image artefacts. These artefacts can be avoided by choosing a less artefact prone array and using an optimised setup. Artefacts can also be compensated for in the preprocessing prior to image reconstruction.

**Related Articles:** Signal to noise ratio (SNR), Pinhole collimator

Coded aperture tomography

(Nuclear Medicine) Coded aperture tomography is a technique that was developed to do limited angle tomography with a gamma camera. The method has been the subject of many scientific papers but is not commonly used in the current clinical environment. In the technique, a number of pinholes are drilled in a solid plate that is placed in front of the scintillation crystal of a gamma camera. Depending on the details of the technique, the pinholes can be random, pseudorandom or structured. There is also a very elegant technique in which a Fresnel zone plate is used.

For a short time before single-photon emission computed tomography became common, a seven-pinhole technique that is somewhat related to coded aperture tomography was used for tomographic imaging of the myocardium using Tl-201.

**Related Article:** Single-photon emission computed tomography (SPECT)
Coded excitation

(Ultrasound) Coded excitation is a generic term for strategies to increase the signal to noise ratio (SNR) in imaging or Doppler modalities. In general, increasing the emitted energy can increase the SNR. There are, however, limits to the energies that can be emitted, as peak intensities are limited by various regulations for safety reasons. In addition, increasing the peak intensities will result in increased non-linear propagation where energy is lost to higher harmonics that are quickly attenuated. A possible remedy is to emit longer pulses coded in a binary form or as a frequency-modulated chirp. As the shape and properties of the transmitted pulse are known, correlation-based methods can be used to extract spatial scattering information. In radar systems, the improvement in SNR can be a factor of thousand or more, but is more modest in ultrasound systems. Examples of such techniques that have been suggested for medical ultrasound are chirps or binary codes such as Golay codes.

Coherence

(General) Coherence manifests itself in the interference between waves. The interference is stationary, i.e. temporally and spatially constant, and depends on the properties of the involved waves. Waves interfere in either a constructive (add) or destructive (subtract) manner depending on their phase relative to one another.

Coherent scattering

(Radiation Protection) Coherent scattering refers to an interaction between a photon and an atom by which the direction of the incident photon is changed, but its energy remains the same. Although in the interaction energy is initially transferred to the atom, it is subsequently given back to the emerging photon, even though the direction of the photon has changed. This process may also be called elastic scattering.

Related Article: Elastic scattering

Coherent source

(General) In physics, coherent source refers to a source that radiates photons with identical phase and frequency, meaning there is no phase shift. The most common use of coherent sources is in the fields of lasers.

Coincidence imaging

(Nuclear Medicine) Coincidence imaging describes an imaging scenario with two opposed detectors providing near-simultaneous detection of two detection events, most notably the annihilation photons resulting from positron emission. Typically, the events fall within 0.5–20 ns of each other. The coincidence data are used to position the annihilation event during the process of image reconstruction.

Coincidence summing of pulse-height spectrum

(Nuclear Medicine) Measurements of radiation from radionuclides that emit multiple photons may sometimes result in events that are incorrect. In some decay schemes, multiple photons can be emitted in cascade (or promptly) because they are part of the same decay. If two or more photons reach the detector and interact, then it is impossible to distinguish between them, leading to a summation of the imparted energy. Examples of such radionuclides are 111-In (emitting 245 and 172 keV) and 60-Co (1330 and 1170 keV). Coincidence summing can also appear in PET systems where an event is registered if two annihilation photons strike two detectors at the same time. The assumption is then that the annihilation has occurred along a straight line defined by the two detectors. Recent PET systems try to take into account the time difference between the two photons due to the distance from the centre of the system as...
information to construct images of less noise. Although these two photons come from the same decay, they arrive to the detectors at different times because of the different distances.

**Coincidence timing window**
*(Nuclear Medicine)* After annihilation, two photons are emitted in almost opposite directions. These two photons will arrive near-simultaneously to their opposite detector. The time difference between their detection will be a function of the distance from the centre of rotation (COR). When an event is registered in a detector, the coincidence processor expects a registration in one of the opposite detectors. Since the speed of light is known and constant, it is possible to calculate the width of the time interval in which the corresponding photon should be expected. The maximum expected time interval is called the coincidence timing window, which typically is 6–20 ns. In order to register a coincidence, there must be two recorded events; one in a detector, which will open a time window and another event in an opposite detector within the time window.

If the count rates are too high, i.e. high administrated dose, there is an imminent risk that there are two or more events registered in an opposite detector within the coincidence timing window. The coincidence processor might have difficulties in finding the true event and therefore misplacing the LOR. These coincidences are called random coincidences (more information in separate article *Event type in PET*).

**Related Article:** Event type in PET


**Cold spot**
*(Radiotherapy)* See Hot and cold spots

**Collection efficiency**
*(Radiation Protection)* When a gas-filled ion chamber is irradiated, the gas in the chamber is ionised. The application of an electric field to the electrodes causes the movement of the charged particles (electrons and ions) towards the electrodes. As the voltage difference between the electrodes is increased from zero to a high value, the current collected increases with voltage until reaching an asymptotic value – the saturation current for the given radiation value, the current collected increases with voltage until reaching a saturation current for the given radiation value.

The electric field strength

\[ E = \frac{V}{d} \]

where

- \( V \) is the voltage applied to the electrode
- \( d \) is the distance between an inner and outer electrode

In a cylindrical ionisation chamber, the electric field lines are radial and the strength \( E \) of the electric field depends on the distance \( r \) according to the following expression:

\[ E = \frac{V}{r \times \ln(b/a)} \]

where

- \( V \) is the voltage applied to the electrode
- \( r \) is the distance from the anode (inner electrode)
- \( b \) is the radius of an outer electrode (cathode)
- \( a \) is the radius of the anode

The collection region is an area in which the electric force \( F \) acting to the charge \( q \)

\[ F = q \times E \]

will cause its collection on the suitable electrode. The geometry of the charge collection can be 2\( \pi \)r or 4\( \pi \).

**Related Articles:** Parallel plate ionisation chamber, Cylindrical ionisation chamber


**Collective dose**
*(Radiation Protection)* The collective dose (usually meant to imply the collective effective dose) describes the total risk to a group of individuals.

When a population is exposed to ionising radiation, each individual will receive an individual effective dose. Although statistically
it is possible to ascribe a risk of late stochastic radiation effects to that individual from this dose, it is much more meaningful, especially in epidemiology, to look at the dose and risk to the population as a whole by summing the mean effective dose to any individual over the total number of individuals exposed.

Thus, mathematically, the collective dose \( S \) is the mean effective dose to an individual multiplied by the number of individuals in the group:

\[
S = \sum_i E_i N_i
\]

where

- \( E_i \) is the mean effective dose to an individual
- \( N_i \) is the total number of individuals in the group

The unit of collective dose is man Sieverts (manSv).

Related Article: Effective dose

Collimation

(Radiotherapy) To be clinically useful, radiation beams must be collimated. In a typical modern medical linac, the photon beam collimation is achieved in a number of ways as listed later. For electron beams, electron beam applicators are also used to create a clinically useful beam, in addition to the primary and secondary collimators.

1. Primary collimator
   This defines the largest available circular field size and is a conical opening machined into a tungsten shielding block, with the sides of the conical opening projecting on to edges of the target on one end of the block and on to the flattening filter on the other end.

2. Secondary movable beam-defining collimators
   These consist of four blocks, two forming the upper and two forming the lower jaws of the collimator. They can provide rectangular or square fields at the linac isocentre, with sides on the order of few millimetres up to 40 cm. Modern linacs incorporate independent (asymmetric) jaws that can provide asymmetric fields, most commonly one-half or three quarter blocked fields in which one or two beam edges, respectively, are coincident with the beam central axis.

3. Multileaf collimator (MLC)
   These are a relatively recent addition to linac dose delivery technology. In principle, the idea behind a multileaf (multi-element) collimator (MLC) is simple; however, building a reliable MLC system presents a substantial technological challenge. The number of leaves in commercial MLCs is steadily increasing, and models with 120 leaves (60 pairs) covering fields up to \( 40 \times 40 \text{ cm}^2 \) and requiring 120 individually computer-controlled motors and control circuits are currently available. MLCs are becoming invaluable in supplying intensity-modulated fields in conformal radiotherapy, either in the step and shoot mode or in a continuous dynamic mode. The MLC may be attached to the linear accelerator as a tertiary collimator or may be incorporated into one set of the secondary collimators.

Collimator Scatter Factor: Radiation is scattered from the collimators. In the case of the beam-defining collimators, this varies with field size because as the collimators open, they present more surface area to the beam, which leads to more scattered radiation. This is why output measured without a scattering phantom, sometimes referred to as \( \text{in-air} \), increases with field size. This factor can be determined by measuring the change in output versus field size \( \text{‘in-air’} \) with a build-up cap large enough to provide electronic equilibrium. These values are normalised to a \( 10 \times 10 \) field.

Collimator Rotation Angle: Some collimators can be rotated in order to provide a better fit for the radiation field to the area of treatment. The degree of rotation is defined by the collimator rotation angle.

Related Articles:
- Multileaf collimator, Electron applicator

Further Readings:

Collimator

(Nuclear Medicine) A collimator is a device that filters gamma and x-rays so that only photons with a direction perpendicular to the collimator surface may pass through. By using a collimator, it is possible to acquire images that represent the activity distribution since the signal recorded in a detector element is assumed to have originated somewhere along a line perpendicular to the detector element (Figure C.48).

The shape, width and length of these holes determine the collimator efficiency and collimator spatial resolution. The purpose of the collimator is to allow photons with a certain direction to reach the detector and at the same time stop other photons that deviate from this direction.

A collimator can be designed to fulfill a certain purpose, i.e. a pinhole collimator is used to get a magnified image of small organs.

**FIGURE C.48** The collimator filters out every photon that has an oblique angle of incidence.
For further details about different collimator designs, see Collimator design.

**Related Articles:** Collimator design, Spatial resolution SPECT, SPECT

**Collimator**

*(Radiotherapy)* See Collimation

**Collimator design**

*(Nuclear Medicine)* Four main types of collimators are used in obtaining the desired image resolution using a gamma camera. These are as follows:

1. Pinhole
2. Parallel-hole
3. Diverging
4. Converging

The way these four collimators are constructed can be seen in Figure C.49. FOV is the field of view and S signifies the direction of the scan.

**Pinhole Collimator:** Pinhole collimators produce an inverted, magnified image. The pinhole collimator is a small opening (a few mm) in a high absorbing metal and is placed at the end of a metal cone, typically 20–25 cm from the detector.

The image is magnified when the distance between the detector and the face of the collimator, \( f \), is greater than the source to detector distance, \( b \), so the source must be placed close to the collimator to get a magnification. When placed close to the collimator, the source is magnified but the camera can only attain a small imaged area. Because of the characteristics of the pinhole collimator, it is used for the magnification of small organs (e.g. thyroid and heart) and for small animal imaging.

**Parallel-Hole Collimator:** The parallel-hole collimator is the most commonly used collimator for gamma cameras and it is described by Cherry et al. (2003) as the ‘workhorse’ collimator. The collimator consists of a number of parallel holes separated by lead septa. The septal walls prevent photons from crossing over from one hole to another.

**Diverging Collimator:** The holes in a diverging collimator diverge from a point typically 40–50 cm behind the collimator. A camera with a diverging collimator produces a minified, non-inverted image. Using such a collimator makes it possible to project sources distributed over a large area onto the detector.

**Converging Collimator:** This collimator can project the magnified images of the source onto the detector when the source is located between the convergence point and twice the length of the collimator. The holes typically converge at a point 40–50 cm in front of the collimator. These collimators can be used to maximise the utilisation for a camera with a large detector area when imaging a small organ.


**Related Articles:** Collimator parameters, Spatial resolution SPECT, SPECT

**Collimator rotation angle**

*(Radiotherapy)* See Collimation

**Collimator scatter factor**

*(Radiotherapy)* Radiation scattered from the collimators makes a significant contribution to the total dose, and depends on field size \( A \) and the collimator scatter factor. In the case of the beam-defining collimators, this varies with field size because as the collimators open, they present more surface area to the beam, which leads to more scattered radiation. This is why output measured without a scattering phantom, sometimes referred to as *in-air*, increases with field size.

The collimator scatter factor, \( S_c \), is the ratio of the output ‘in-air’ for a given field size to that for the reference field:

\[
S_c(A,E) = \frac{X(A,E)}{X(10,E)} = \frac{D(A,E)}{D(10,E)}
\]

**FIGURE C.49** The four most common collimator designs used in emission tomography.
This factor can be determined by measuring the change in output versus field size ‘in-air’ with an ionisation chamber plus a build-up cap large enough to provide electronic equilibrium. These values are normalised to a 10 × 10 field. The measurement set-up is shown in Figure C.50.

This is otherwise known as the collimator factor (CF), or the air output factor, or the relative exposure factor (REF).

**Related Articles:** Peak scatter factor, Scatter factor, Tissue–air ratio, Collimation

**Collision kerma (Radiotherapy)** A major part of the initial kinetic energy of electrons in low-atomic-number materials (e.g. air, water, soft tissue) is expended by inelastic collisions (ionisation and excitation) with atomic electrons. Only a small part is expended in the radiative collisions with atomic nuclei (Bremsstrahlung, electron–positron annihilation). The total kerma ($K$) can thus be divided into two parts: the collision kerma ($K_{\text{coll}}$) and the radiative (radiation) kerma ($K_{\text{rad}}$):

$$K = K_{\text{coll}} + K_{\text{rad}}$$

The fraction of energy lost through the radiative processes is known as the Bremsstrahlung fraction ($g$) and therefore the fraction due to collision interactions is $(1 - g)$. So it is possible to relate the collision kerma to the total kerma in the following expression:

$$K_{\text{coll}} = K (1 - g)$$

For monoenergetic photons, the relationship between the collision kerma ($K_{\text{coll}}$) and the photon energy fluence ($\Psi$) at that point in the medium is given by

$$K_{\text{coll}} = \Psi \left( \frac{\mu_{\text{en}}}{\rho} \right)$$

where ($\mu_{\text{en}}/\rho$) is the mass energy absorption coefficient.

For polyenergetic beams, the collision kerma can be expressed as a relationship between the photon energy fluence spectrum $\Psi_k(E)$ and the mass energy absorption coefficient as follows:

$$K_{\text{coll}} = \int \Psi_k(E) \left( \frac{\mu_{\text{en}}}{\rho} \right) dE$$

that is

$$K_{\text{coll}} = \Psi \left( \frac{\mu_{\text{en}}}{\rho} \right)$$

where

$\Psi$ is the photon energy fluence (integral distribution)

($\mu_{\text{en}}/\rho$) is the mass energy absorption coefficient averaged over all photon energies, weighted by the spectral distribution of the photon energy fluence

It is clear that while the kerma occurs at a particular point, the absorbed dose (absorption of energy by the material) will take place at a different location due to the finite range of the secondary electrons produced. The collision kerma can be related to the absorbed dose, since the radiative photons will mostly escape the volume of interest, by the following approximation:

$$\beta = \frac{D}{K_{\text{col}}}$$

By assuming all radiative photons escape the volume of interest, then $\beta$ is approximately 1. In the build-up region of a x-ray photon beam, $\beta$ is less than 1, at the depth of maximum dose, $d_{\text{max}}$ (charged particle equilibrium) $\beta$ is equal to 1, and at depths beyond this point (transient charged-particle equilibrium), $\beta$ is greater than 1. The relationship between collision kerma and absorbed dose at depths beyond $d_{\text{max}}$ is almost constant because the average energy of electrons generated does not change noticeably with increasing depth. This is illustrated in Figure C.51.

Collision kerma may also be expressed in respect to exposure by the following equation:

$$(K_{\text{coll}})_{\text{air}} = X \left( \frac{W_{\text{air}}}{e} \right)$$

where

$W_{\text{air}}/e$ is the average energy expended in air per ion pair formed (the current most accurate value for this is 33.97 J/C)

$X$ is exposure (units C/Kg) given by the following relation:

$$X = \frac{dQ}{dm}$$

$\beta$ is the factor by which the absorbed dose is multiplied by the collision kerma.

**FIGURE C.50** Set up for measuring the collimator scatter factor.
where \( d\Omega \) is the absolute value of the total change of ions of one sign produced in air when all the electrons and positrons produced by photons in mass \( dm \) of air are completely stopped in air.

**Related Articles:** Kerma, Air Kerma, Exposure, Bremsstrahlung


**Collision mass stopping power**

*(Radiation Protection)* Mass stopping power is defined as the change (loss) in energy of the particle per unit length of a particle’s path, i.e. \( dE/dx \). The total mass stopping power \( (S_{\text{tot}}) \) is equal to the sum of the collision mass stopping power \( (S_{\text{coll}}) \) and the radiation mass stopping power \( (S_{\text{rad}}) \):

\[
S_{\text{tot}} = S_{\text{coll}} + S_{\text{rad}}
\]

(C.3)

The interaction of electrons with the electric field of the nucleus to produce Bremsstrahlung gives rise to radiation mass stopping power. This process is more significant at high electron energies and for media of high atomic number.

The collision mass stopping power itself is the change in energy per unit path length due to elastic collisions with the atomic electrons of a medium. Collision energy losses decrease very slowly in materials with atomic number, due to the greater binding energy possessed by these materials. Collisions with atomic electrons cause excitation and ionisation of the atoms within the medium that a certain particle is travelling through. Where the collisions lead to the total energy of the incident particle being absorbed at the point of the interaction, or close by without the energy being transferred away, then it is referred to as the restricted collisional mass stopping power. This concept is the most useful at a microscopic level in describing how energy is deposited and leads to absorbed dose.

The collision mass stopping power is equal to the sum of the mass stopping power associated with soft collisions \( (S_{\text{sc}}) \) and that associated with hard collisions \( (S_{\text{hc}}) \):

\[
S_{\text{coll}} = S_{\text{sc}} + S_{\text{hc}}
\]

(C.4)

**FIGURE C.51** Relation between absorbed dose and collision kerma for a megavoltage photon beam. The point designated as c.e.p. is the centre of electron production. (After Loevinger, R., *Med. Phys.*, 8, 1981.)

Soft collisions account for around 50% of all collisions. They occur when an outer electron is dragged off by a passing charged particle, causing excitation and ionisation. The excitations that result from soft collisions give rise to low-level energy transitions and associated Čerenkov radiation. Hard collisions result from interaction between a charged particle and a single electron. This interaction may lead to the ejection of a delta ray (high-energy electron) and characteristic x-rays (Figure C.52).

Total mass stopping power varies for electrons and protons. Electrons lose energy steadily and have a definite range. For electrons, change in energy varies more significantly with length travelled (depth in tissue). Protons exhibit a Bragg peak, where they react more significantly. This effect is used to the physicist’s advantage in radiotherapy: treatments are planned so that the proton Bragg peak, and its increased reaction rate, coincides with the treatment area (Figure C.53).

**Related Articles:** Stopping power, Linear stopping power, Bethe–Bloch equation, Bremsstrahlung, Delta ray, Čerenkov radiation

**Collision sensor**

*(Nuclear Medicine)* This is a sensor used to prevent heavy camera equipment hitting a patient or other equipment. The sensors can be placed on the mobile parts of the detector. One example is the collimator collision sensor on a gamma camera, which registers when the collimator strikes an object and stops the motion, preventing the camera moving too close to the patient.

**Collision sensor**

*(Radiotherapy)* As a result of the ability of various components of a linac to move independently (gantry, couch, EPID), there is a possibility that a patient, or indeed an operator of the equipment, could be crushed, and equally there is a chance of a collision between the moveable components, which could result in significant (expensive) damage. The reasons for such an incident could be due to either incorrect operation of the equipment or simply electric failure of one or more controlling motors, causing them not to stop when required.

**FIGURE C.52** Soft and hard collisions.

**FIGURE C.53** Electron and proton mass stopping power ranges: (a) electron mass stopping power and (b) proton mass stopping power.
Therefore, it is important for the safety of both personnel and equipment that some means of avoiding, or at least minimising the damage, is in place. The simplest collision sensor employed is a ‘touch ring’ located externally on the head of the gantry (and also employed on some EPID arm systems). The ring contains a micro-switch held in the closed position such that when a tiny deflection in the ring occurs, the switch opens and interrupts the power supply to all drive motors. The operator must then reset the switch to the closed position before continuing with any motion. Similar techniques using inductive proximity sensing circuits are also used.

An alternative solution employed in some departments is to use simple software to calculate combinations that would potentially cause a collision or crush a patient. The user enters the values and can quickly confirm whether such a situation would occur. This solution removes the necessity of an external ‘touch ring’, but it does not take into account the variability in size and position of patients and any other additional objects placed in the room.

Related Articles: Interlock, Interlocking device

Collisional energy loss (Radiation Protection) Collisional energy loss occurs as a result of soft and hard collisions between particle radiation, or the secondary electrons caused by ionising radiation, and atoms within the absorber. For a definition of these interactions, see Collision mass stopping power.

Related Articles: Secondary electrons, Secondary ionisations, Collision mass stopping power

Colour Doppler (Ultrasonography) The common shorthand for colour flow imaging where estimates of the mean velocity are overlaid onto the B-mode image as colour-coded pixels. The mean velocity is found from an ensemble of pulses (4–16) and the mean phase shift that has occurred between them, by what is referred to as the auto-correlator approach (see Autocorrelation). A more extensive description of the technique can be found under the article Colour flow imaging.

Colour flow imaging (CFI) (Ultrasonography) Colour flow imaging (CFI) conventionally refers to the depiction of a colour-coded map of flow superimposed on a B-mode image. Colour is used to depict areas of the image in which movement is detected by ‘Doppler’ methods. Commercial colour flow ultrasound scanners were introduced in the late 1980s when autocorrelation methods were used to produce 2D colour flow maps at clinically useful frame rates and spatial resolutions.

The image (Figure C.54) shows a conventional colour flow image of flow in renal vessels. Colour is usually portrayed as red hues in one direction relative to the beam, blue hues in the opposite direction, but other colour maps may be used. Change in hue, for example from red to orange, represents changes in velocity vector and is in turn dependent on the speed and direction of the blood relative to the beam. The colour image depicts small vessels, which may not be evident on B-mode because of their small size. Because colour flow signals are highest when flow is aligned to the ultrasound beam, whilst in B-mode, contrast is best when structures are orthogonal to the beam. Colour can be used to identify vessels for pulsed-wave Doppler investigation.

The colour flow image is very dependent on instrument factors, including pulse repetition frequency, gain, wall filter, transmit frequency, size of the area under investigation (colour area), line density used and the ensemble length. The need for separate pulses for colour reduces frame rate of the ultrasound image and the trade-offs between frame rate, colour spatial resolution and sensitivity are complex. These factors are partly under operator control and partly optimised by the scanner in order to ensure an acceptable image. As the colour map is, in effect, a separate image from the B-mode, there are compromises made in depicting both in the final image, with the possibility of one scan masking information from the other.

Narrowband and wideband techniques are now used commercially for CFI. Other CFI displays used are power Doppler, colour M-mode imaging and 4D CFI.

Colour saturation (Diagnostic Radiology) See Hue, saturation, luminance (HSL)

Columnar caesium iodide (Diagnostic Radiology) See Caesium iodide

Comet tail (Ultrasonography) The lung is normally considered poorly accessible using ultrasound because ultrasound is heavily attenuated by air. Air also produces reverberation artefacts under the lung surface. This however does not mean that ultrasound is a poor diagnostic tool for conditions associated with the lung.

The ‘comet tail artefact’ is an echographic image obtained with a cardiac ultrasound probe positioned over the chest and consists of multiple lighter beams ('comet tails') fanning out from the lung’s surface. These ‘comet tails’ originate from water-thickened interlobular septa. The presence and the number of ‘comet tail’ images provide information on interstitial pulmonary edema, making ultrasonography an attractive, easy-to-use, bedside diagnostic tool for assessing extravascular lung water (oedema or pulmonary congestion). The ‘comet tail’ artefact is a well-known and widely used marker of pulmonary oedema.

Comforters and carers (Radiation Protection) ‘Comforter and carer’ is defined in Regulation 2(1) of the Ionising Radiation Regulations 1999 (IRR 99) as an individual who knowingly and willingly incurs an exposure to ionising radiation resulting from the support and comfort of another person who is undergoing or who has undergone any medical exposure.

When a person attending with a patient who is a relative or friend is expected to receive an exposure to ionising radiation, whether at the time of the imaging investigation or radiation therapy, or as a consequence of tending to the patient afterwards, he or she may receive...
a radiation dose in excess of the annual public dose limit (1 mSv). In such cases, the person should be designated as a comforter and carer. Comforters and carers are not subject to dose limits and can enter controlled areas. However, the radiation exposure to the comforter and carer should only be conducted in accordance with suitable written arrangements designed to restrict radiation exposures to previously established constraints and so far as is reasonably practicable. A reasonable constraint used is to limit the dose to the comforter and carer to no more than 5 mSv in any 5 year period.

It is essential in designating a person as a comforter and carer to ensure that they are given sufficient information and instruction about the hazards and risks and that they therefore give informed consent.

**Commissioning**

(Radiation Protection) Commissioning is a term to describe the testing processes required to bring a new piece of equipment into clinical use. Commissioning involves familiarisation with new equipment, in order to understand how to undertake the routine procedures (both clinical and non-clinical) for which it has been purchased.

Some aspects of commissioning may be undertaken at acceptance testing, for example baseline values will be determined for comparing with future quality control checks. The clinical commissioning should begin with applications training, where specialists employed through the supplier train local operators on equipment use, and lead to a comprehensive understanding of the equipment’s uses and limitations. Operating procedures should be written to describe all the necessary operational functions and features of the new equipment in order for it to be brought safely and effectively into clinical use.

**Committed dose**

(Radiation Protection) Following intake to the body of a radioactive substance, biokinetic models describe the uptake and expulsion of the radioactive substance over time in individual organs and tissues. It is necessary to know both the radionuclide and the chemical form to carry out this biokinetic analysis. The combination of such information, together with the physical decay (half-life) of the substance can then be used to determine the radiation dose received by the person over the time that the substance remains in the body. This is called the committed dose to the person, defined by the International Commission on Radiological Protection (1991).

For each tissue or organ, there will be a committed equivalent dose $H_T(t)$ based on the activity accumulated within the tissue, the type(s) of radiation emitted by the substance, and the time it remains there, known as the integration time $t$. If this time is not specified for particular radioactive substance and pathway of intake, then it is taken to be 50 years for adult intake, or 70 years for children.

If the committed equivalent doses are multiplied by the relevant tissue weighting factors and summed for the whole body, then we have an analogous quantity to effective dose − i.e. committed effective dose $E(t)$.

**Related Articles:** Equivalent dose, Effective dose, Radiation weighting factor, Tissue weighting factor


**Compartment**

(Nuclear Medicine) A compartment is a space in which substances or tracers are distributed uniformly and in a specified form or forms. The rate at which substance/tracer is transported out of a compartment is proportional to the amount in the compartment.

A tracer compartmental model is a mathematical description of the transport/reaction pathways of a tracer in terms of interconnected compartments.


**Compartment models**

(Nuclear Medicine) A compartment is defined by a volume or space within which the tracer is rapidly distributed uniformly over the volume, which means that there are no significant concentration gradients. The spatial extent of a compartment is sometimes defined by clear physical boundaries, for example the intravascular blood pool, reactants and products in a chemical reaction and substances that are separated by a membrane. Other compartments have different restrictions, for example the tracer can be metabolised or trapped in one of two different cell populations in an organ, and thus the two cell populations represent two different compartments.

Additionally, two different radioisotopes do not necessarily share the same compartments. For example, labelled red blood cells are distributed intravascularly while thallium is distributed both intravascularly and extravascularly. Therefore, a compartment model (i.e. number, interrelationships, organisation and definition of compartments) must be calculated for each tracer.

**Related Articles:** Tracers, Analogue tracers


**Compensating filters**

(Radiotherapy) A compensating filter can be positioned in the path of the beam to modify the dose distribution. These filters are typically used for more complex situations where a simple wedge or an additional layer of tissue equivalent material (bolus) is not suitable.

The filters are manufactured from a high-density material and positioned on an accessory tray in the head of the linac. The filters are extremely time-consuming to produce for each individual field and patient. They typically consist of small blocks of material of varying thickness across the whole field.

The availability of IMRT with MLC results in the reduction of the use of such filters.

**Abbreviations:** IMRT = Intensity-modulated radiation therapy and MLC = Multileaf collimator

**Related Articles:** Compensation, Compensator, Wedge filter, Bolus


**Compensating wedge**

(Radiotherapy) See Wedge

**Compensation**

(Radiotherapy) The term compensation refers to the process of modifying a dose distribution to take account of an oblique patient surface across the field. This is achieved by placing a compensator (compensating filter, wedge filter, bolus) in the beam path.

IMRT can also be said to provide a form of compensation.

**Abbreviation:** IMRT = Intensity-modulated radiation therapy

**Related Articles:** Compensator, Compensating filter, Bolus, Wedge filter
Compensator
(Radiotherapy) A compensator is used to modify the dose distribution whenever the beam is incident on a non-uniform, or oblique, patient surface. For simple geometric situations, a compensator can be a wedge filter or an additional layer of tissue equivalent material (bolus). However, for more complex problems, a compensating filter must be uniquely made for each field.

IMRT can also be said to provide a form of compensation.

Abbreviation: IMRT = Intensity modulated radiation therapy.

Related Articles: Compensating filter, Compensation, Bolus, Wedge filter

Complex number
(Nuclear Medicine) A complex number, z, is an extension of the real number system and can be expressed in the form of

\[ z = x + yi \]

where \( i \) is defined as

\[ i^2 = -1 \]

where

\( y \) and \( x \) are real numbers
\( yi \) and \( x \) are called the imaginary and real part of the complex number, respectively

One of the useful properties of complex numbers is that one can obtain negative real numbers by squaring complex numbers. Alternatives to the Cartesian representation of a complex number are the polar or exponential representation

\[ z = re^{i\theta} = r \cos \theta + ir \sin \theta \]

where

\[ r = |z| = \sqrt{x^2 + y^2} \]
\[ \theta = \text{arg}(z) = \tan^{-1} \left( \frac{y}{x} \right) \]

where

\( r \) is called the modulus of \( z \)
\( \theta \) is called the argument or phase of \( z \)

Complication-free tumour control
(Radiotherapy) The aim of radiotherapy is to achieve complication-free tumour control, i.e. deliver enough radiation to the tumour to destroy it without irradiating normal tissue to a dose that will lead to serious complications. For more information on the effectiveness of radiotherapy treatment, see the article on Therapeutic effect.

Related Article: Therapeutic effect

Composite
(Nuclear Medicine) A composite material refers to a substance consisting of two or more materials. The two materials have different physical properties and remain separate on a macroscopic level. Composite materials can be made to be both lightweight and strong. Composites also offer a low attenuating alternative for patient couches used in computed tomography and different emission tomography modalities.

Related Article: Couch (patient)

Composite transducer
(Ultrasound) PZT, lead zirconate titanate, is the most commonly used transducer material. This material is a synthetic ceramic, which can be produced in various versions for different applications requiring particular properties. However, the mismatch in acoustic impedance between PZT (3.0 × 10^7 kg/m²/s) and soft tissue (1.6 × 10^6 kg/m²/s) is a significant disadvantage for the transmission of ultrasound energy into tissue and for the ability of transducers to produce short acoustic pulses necessary for high-resolution imaging. The mismatch will cause internal reverberations within the transducer element.

One way of reducing this mismatch is to use a composite transducer where the PZT elements are embedded in a polymer matrix with lower acoustic impedance. The advantages of a composite transducer are better matched impedance, broad bandwidth and high sensitivity.

Related Articles: Transducer, Backing material, Matching layer, PZT


Compound filter
(Radiotherapy) In the kilovolt range, radiotherapy is normally delivered using metal filters positioned in the beam to modify the radiation quality. In this way, softer (lower energy) x-ray photons that would irradiate the superficial layers of the skin and would not contribute to the dose at the desired point of delivery are removed from the beam.

A compound or composite filter uses two or more materials that complement each other in their absorbing abilities and are constructed with higher atomic number (Z number) closest to the tube and with lower atomic number closest to the patient.

As the beam passes through the filter, most filtration occurs in the higher atomic number material and the lower atomic number second layer of the filter absorbs any characteristic radiation emerging from the first. With a copper and aluminium compound filter, photoelectric attenuation in copper produces characteristic radiation with an energy of about 8 keV; this would be energetic enough to reach the patient and increase skin doses. However, the aluminium positioned on the patient side of the copper absorbs this characteristic radiation. Aluminium’s own characteristic radiation has a low energy (1.5 keV), and this is absorbed in the air gap between the patient and filter.

Related Articles: Skin sparing, Kilovoltage (kV)


Compound imaging
(Ultrasound) Compound imaging uses multiple acquired images to produce the final image. In current commercial usage, it is usually applied to images that are a compound of images obtained from different transmit steering angles or frequencies.

Three limitations of ultrasound are as follows:

1. Limited field of view from a single transducer
2. Directional dependence in the image
3. Speckle

Compound imaging has, at times, been used to address all these limitations.

Early compound scans used multiple images from different scan angles to produce a compounded scan. The advantage of this system is that some tissue interfaces are better seen from particular scan direction and that effect of speckle is reduced since the speckle pattern is different from different scan angles and compounding produces some averaging of the speckle in the final image. It is
Compound nucleus

Obviously imperative to have a clear spatial reference for the image, to align and merge the images produced from different views.

The system is shown diagrammatically in Figure C.55. An early system using this approach was the Diasonograph® scanners from Glasgow in the early 1960s. A more recent variation of the technique is ultrasound tomography whereby tissue can be imaged from a circular array of transducers or transducers swept in a circular motion. Compounded images may be obtained from pulse-echo images or transmission ultrasound techniques using speed of sound and attenuation properties. Systems using these techniques have been developed for breast imaging.

Current commercial systems can use spatial compounding and frequency compounding to obtain images from a single transducer using different transit directions (Figure C.56) or different frequencies. This offers the advantage of reducing speckle artefact and improving echoes from interfaces over a wider angle than with conventional transmit paths. Since each compounded image requires more transmit pulses, effective frame rate is reduced. If artefacts are part of the diagnostic pattern recognition, for example in shadowing from microcalcifications, then the use of spatial compounding may not be recommended since it will inherently reduce this effect.


In Figures C.57 and C.58, the effect of spatial compounding is demonstrated.

**Compound nucleus**

(*Nuclear Medicine*) A compound nucleus is formed in a reaction in which two nuclei, or a bombarding particle (e.g. a neutron) and a target nucleus, combine into an excited nucleus. The nucleus lives for a certain time, and the nucleus ‘forgets’ how it was formed. It decays by ‘evaporation’ of nucleons from the compound nucleus, and by emission of gamma photons. In the case where the bombarding particle is a neutron, the process may be illustrated by

\[ n + ZA X + X^* \rightarrow ZA X^* \rightarrow ZA X + \gamma \]

where

- \( ZA X \) is the target nucleus
- \( ZA X^* \) is the unstable compound nucleus
- \( ZA X^* \) is the excited nucleus

**Further Reading:** Podgorsak, E. B. 2006. Radiation Physics for Medical Physicists, Springer, Berlin, Germany.
**Compound scan**  
(Ultrasound) A scan where individual images obtained from different beam directions or at different frequencies are combined to produce the final image.  
**Related Article:** Compound imaging

**Compounding**  
(Ultrasound) Compounding is the process by which individual images from different beam directions or frequencies are combined to produce a final image.  
**Related Article:** Compound imaging

**Compressibility**  
(Ultrasound) Compressibility ($\kappa$) is the relative volume change of a fluid as a response to pressure change:  
$$\kappa = -\frac{1}{V} \frac{\partial V}{\partial p}$$

The compressibility together with density determines the sound speed in a fluid.

**Compression**  
(Ultrasound) The word compression is used in two different ways in diagnostic ultrasound. For compression of the medium by the ultrasound wave, see *Rarefaction*.

Electrical signals can also be compressed. Compression of the dynamic range of the received echo signals in a B-mode scanner is of considerable importance. The received signals can have amplitudes in a 60 dB range, from $\mu V$ to V while images in the display use a 20 dB range of brightness levels. The transformation from a 60 dB range to a 20 dB range is effected by non-linear amplification where the smaller signals are amplified more than the larger signals. This is called compression.

Different applications acquire different degree of compression. This is why many scanners have a control to adjust the non-linear amplification curve, often labelled 'compression' or 'dynamic range'.  
**Related Articles:** Rarefaction, Dynamic range

**Compton effect**  
(Radiation Protection) Also known as Compton scatter, incoherent scatter or inelastic scatter.

When a photon strikes an outer orbital or unbound electron, there is some transfer of energy to the electron, but the photon is not completely stopped. The process may be thought of in terms of a billiard-ball type collision. (Figure C.59) The incident photon interacts with a ‘free’ (outer shell) electron. The incident photon strikes the electron whence the electron and the scattered photon are emitted in separate directions. Some of the photon energy is transferred to the electron and the energy lost by the incident photon increases as the scattering angle increases (i.e. the angle between the initial direction of the photon and the direction of the scattered photon).

The Compton effect dominates interaction processes at medium energies (90 keV to 2 MeV in water or human tissues; between 200 keV and 2 MeV, the only interaction is Compton). The process is also known as inelastic scatter. The ejected electron is known as a Compton electron or recoil electron.

The likelihood of interaction for the Compton effect varies with $1/E$ where $E$ is the energy of the incident photon, and with electron density, represented by $N_e Z / A$ where $N_e$ is Avogadro’s constant representing the number of atoms per unit mass, $Z$ is atomic number and $A$ is atomic mass. Hence, Compton effect is relatively independent of the $Z$ of the material.

![Illustration of Compton effect](image)

**Compton electron**  
(Radiation Protection) An electron ejected from an atom as a result of a Compton interaction with an incident photon. Only some of the energy of the photon is given to the electron, and the photon is scattered.  
**Related Article:** Compton effect

**Compton interaction**  
(Radiation Protection) Interaction between an incident photon and an outer shell (loosely bound) atomic electron in which only some of the energy of the photon is transferred to the electron. The photon is scattered to a new direction, and the electron is ejected from the atom with kinetic energy equal to the loss of energy of the photon, minus the work function for ejection of the electron.

Also known as inelastic scatter, incoherent scatter, or Compton scatter.  
**Related Article:** Compton effect

**Compton scattering**  
(Radiation Protection) Interaction between an incident photon and a loosely bound atomic electron in which only some of the energy of the photon is transferred to the electron. The photon is scattered into a new direction, and the electron referred to as Compton or recoil electron is ejected from the atom with kinetic energy equal to the loss of energy of the photon.  
Also known as inelastic scattering, incoherent scattering or Compton interaction.  
**Related Article:** Compton effect

**Computed radiography (CR)**  
(Diagnostic Radiology) Computed radiography (CR) is an x-ray diagnostic radiology imaging process that uses photostimulable phosphor plates. In most cases, the CR plates are a direct replacement for film-screen cassettes, so conventional imaging equipment can be used. In the process,  
- A plate is inserted into the cassette holder and exposed in the conventional way.  
- The plate is taken to a reader and read out using laser stimulation.  
- The plate is erased using bright white light.
The main advantage of CR is that the dynamic range is much greater than screen film. Thus, errors in exposure do not cause as great a degradation of image quality. Resolution is less than screen film, but the values are acceptable in most imaging situations.

CR systems use the existing x-ray tube, generator and stand. The only modification of the x-ray system is that the x-ray film cassette is replaced by a storage phosphor cassette. This makes the transfer from classical radiography to CR very cost effective.

Some sources use the term computed radiography (CR) instead of the term digital radiography (DR), or vice versa. These two terms differ as CR and DR use different detectors (Figure C.60). CR systems use photostimulable phosphor plates as the detector. These systems were developed before the DR systems and used the phosphor plate as a replacement of the x-ray film. Direct DR uses x-rays transferred to charge by amorphous selenium detectors, and indirect DR uses phosphor and photodiode or CCD detector. See the specific articles for these systems.

Related Article: Storage phosphor

**Computed tomography (CT)**

*(Diagnostic Radiology)* Computed tomography, usually referred to as CT, is a medical diagnostic imaging technique, which uses x-rays to create cross-sectional images, or ‘slices’, of a patient’s anatomy. The medical CT scanner was the invention of British engineer, Sir Godfrey Hounsfield, and the first clinical scans of a patient’s head were obtained in 1972 at Atkinson Morley’s Hospital in Wimbledon, London (Figure C.61). Later, the technique was refined for body imaging. In 1979, Sir Godfrey, and American scientist Allen Cormack, were awarded the Nobel Prize in Medicine for the invention.

The basic imaging principle of CT is the same as that of plane-film radiography, in that it relies on the differences in the x-ray attenuating properties between materials. CT does not use radiographic film to acquire the image information, but digital signals from the detectors are sent to a computer for processing and then displayed on a monitor.

The principal components of a CT scanner are a collimated source of radiation, usually a diagnostic x-ray tube, a radiation detector and a computer. The x-rays emitted pass through the object to be imaged, and the detectors measure those transmitted. The x-ray source and detector are rotated through a minimum of 180° around the object, and attenuation information is obtained from different angles, or views. The data are processed by the computer, enabling a cross-sectional image of the object to be reconstructed. Currently, most CT scanners use filtered back-projection for image reconstruction. However, iterative reconstruction methods are being implemented on the latest models of scanners.

Early CT scanners were of the so-called first-generation geometry (Figure C.62). An x-ray pencil beam and single detector translate across the field of view to acquire an attenuation profile. The assembly (x-ray beam–detector) then rotates through small angles, each time acquiring a further profile.

The schematic diagram shows a third-generation CT scanner (Figure C.63), in (a) the scan (x–y) plane and (b) the sagittal (y–z) plane. This design is now almost universally employed in medical CT scanning. It employs an x-ray fan beam in the x–y plane, and an arc of detectors, whose extent is generally large enough to encompass the whole object being imaged, eliminating the need for translation. The tube and detector assembly rotate around the z-axis to acquire data for image reconstruction. Along the z-axis, the x-ray beam is collimated, so that information acquired in a single rotation is limited to a narrow ‘slice’ of tissue. In recent years, the multislice CT scanner has been introduced, which enables the simultaneous acquisition of a number of ‘slices’ in a single rotation.

**Related Articles:** CT reconstruction, Multislice CT scanner, Helical scanning

**Computer-aided diagnosis (CAD)**

*(General)* Computer-aided diagnosis is the application of data processing to clinical data and clinical images, with each patient’s data
being analysed by a computer, which offers a tentative diagnosis or range of diagnoses with their associated statistical probabilities.

**Computer-aided perception (CAP)**

*(General)* Computer-aided perception is the use of image processing of clinical data to assist clinicians by providing additional visual clues to aid in the diagnosis and treatment of patients.

A basic form of aided perception would be to use image processing techniques to highlight areas of mammograms, which the computer calculates as abnormal, and highlight these automatically for the reporting radiologist.

More complex processes include the generation of ‘virtual reality space’, based on 3D imaging data such as helical CT or MRI and using sophisticated tissue segmentation algorithms and 3D display software to provide a virtual patient, presenting the information to the clinician in a clearer and more easily comprehended model.

This ‘virtual reality’ may be projected back into the space occupied by the patient, such as in some image-guided therapies, producing additional 3D visual clues as to the subsurface location of tumour or important blood vessel.

**Computer-controlled accelerator**

*(Radiotherapy)*

**Networking:** To achieve the required dose delivery, modern linear accelerators are computer controlled. The main delivery control mechanism is represented by the record and verify system (RV system). Treatment parameters are transferred electronically from the treatment planning system to the RV system. RV system holds the patient treatment parameters and checks them against the parameters used in daily set up to confirm that the set up is correct.

**Computer Control:** In addition, the working configuration and many internal operating parameters of the linear accelerator are set and monitored by the system’s own dedicated computer. These parameters are, for example output, beam flatness and symmetry. If they exceed predetermined tolerances, then software interlocks will switch the linear accelerator off until parameters are adjusted back within tolerance.

**Abbreviation:** RV system = Record and verify system.

**Related Article:** Interlock

**Concave target volume**

*(Radiotherapy)* A concave target volume is defined as one who has at least one interior angle greater than 180°. The opposite of a concave target volume is a convex target volume, see Figure C.64 for theoretical examples.

Concave target volumes can be treated successfully with intensity-modulated radiotherapy, which allows critical structures within the concavity of the target volume to be spared in the dose.
distribution. Typical tumour sites and respective critical structures that require concave target volumes include

- Harynx (spinal cord)
- Prostate (rectum)
- Skull base (optic pathway structures, brainstem)

Concave target volumes cannot be treated successfully with multiple beams with no intensity modulation.

**Related Articles:** Target volume, Conformal radiotherapy, Beam’s eye view, Intensity-modulated radiotherapy


**Concrete**

*(Radiation Protection)* See Radiation shielding

**Condenser chamber**

*(Radiation Protection)* A number of ionisation chambers can be described (in broad terms) as capacitors. These are initially charged to certain potential $V_1$ (prior the measurement). The ionising radiation graduually discharges the chamber and the resulting decrease of the voltage is used to estimate the integrated ionisation charge. The charge liberated by the radiation is

$$Q = C(V_1 - V_2)$$

where

- $C$ is capacity of the condenser chamber
- $V_1$ is the initial charge of the condenser chamber (prior the measurement)
- $V_2$ is the voltage after the ionisation radiation has partially discharged the chamber (at the end of the measurement)

Such condenser chambers use a capacitor connected to their central electrode. This capacitor collects the liberated charge. Later a reading device measures the integrated charge in the capacitor. This design is convenient for measurements where the cable between the chamber and the electrometer has to be avoided as a source of noise. Often condenser ionisation chambers are used to measure low exposure rate, for example as personal pencil dosimeters (integrating dosimeter, pocket dosimeter).

*Figure C.65* shows a condenser chamber consisting of a central electrode and outer electrode (chamber wall). The chamber is filled with air and connected to the electrometer. The chamber is charged and placed in the radiation field for a period of time in which the integrated ionisation charge is measured.

A typical example of condenser chamber is the so-called pocket chamber. It is with small size (like pen) and equipped with a quartz fiber electroscope for reading the dose on an internal scale.

The ionisation condenser chamber can perform as well-type (*Figure C.66*) designed to measure the radioactivity of the source in a $4\pi$ counting geometry. The walls are made of brass or steel and the inner electrode of a thin foil of aluminium or copper. The gas pressure in the chamber can be changed to increase the ion current. This type of condenser chamber is used for radiation source calibration, for example of gamma radiation. The ion current from an unknown source is compared with that measured from a standard one.

**Related Articles:** Integrating dosimeter, Cylindrical ionisation chamber, Ionisation chamber


**Conduction band**

*(General)* The conduction band refers to a range of electron energy, higher than that of the valence band. It is either unfilled or partially filled by conduction electrons (*Figure C.67*).

Specific energy is required for each electron to migrate from the valence band to the conduction band. In case an electron jumps to this higher energy level, it resides there for fractions of a second, then returning back to the valence band by emitting the energy in the form of heat, light, or transferring it to another electron.

Conduction band electrons are free to move around the crystal. Upon application of external electric field, the conduction electrons accelerate.
Conductivity

(General) Conductivity, or specifically electric conductivity, is defined as the ability of a substance to conduct electric current between two points. The conductivity is the reciprocal (inverse) of the more commonly encountered term, resistivity ($\rho$) and depends on the number of charge carriers in the material and their mobility. It depends on the nature of the substance, its cross-sectional area and its temperature. All substances possess conductivity to some degree, but the amount varies widely, ranging from extremely low (insulators such as benzene, glass) to very high (silver, copper and metals in general). Metals are better conductors of electricity because of their high free-electron density, while non-metals, such as rubber, are poor conductors and may be used as electrical insulators. Increasing the cross-sectional area of a given conductor will increase its conductivity since more electrons will be available for conduction. However, increasing the temperature of a metal conductor will reduce its conductivity due to the electrons ‘bumping’ into each other more as they are moving faster. Controversially in a non-metal, an increase in temperature improves conduction because it frees more electrons. Conductivity measurements are widely used in industry for liquids, which generally consists of ionic compounds dissolved in water. These solutions have conductivities approximately midway between insulators and metallic conductors. The conductivity is easily measured by electronic means, and provides information about the quality of the water, or the make-up of the solution. Electrical conductivity is commonly represented with $\sigma$ and has the SI units of siemens per metre (S/m).

Cone

(Diagnostic Radiology) A very simple beam restrictor – a metal extender (cylinder cone) attached to the diaphragm of an x-ray equipment. See article on Beam restrictor.

Related Articles: Beam restrictor, Diaphragm

Cone beam artefact

(Diagnostic Radiology) The cone beam artefact occurs on multislice scanners with extended coverage in the z-direction (see Cone beam CT). For scanners acquiring more than four simultaneous slices, the beam can no longer be assumed to have parallel beam (pencil beam) geometry along the z-axis.

The diverging beam along the z-axis results in inconsistent projection data from opposing views, and this effect increases with distance from the central ray. Figure C.68a shows (in an exaggerated way) the path of the ray to the outermost detector. When the x-ray tube is at $0^\circ$, the ray traverses a different path through the patient in comparison to the situation when the tube is at $180^\circ$. These inconsistencies give rise to the cone beam artefact. The inconsistencies are not present for the central rays in the x-ray beam as the beam follows the same path through the patient in opposing views (Figure C.68b).

Figure C.69 shows images of a Teflon rod angled relative to scan axis: (a) absence of cone beam artefact and (b) appearance of cone beam artefact. (Graphs courtesy of ImPACT, UK, www.impactscan.org)

The artefact is seen in images from the outer detector rows (Figure C.69b). The severity of the artefact increases for objects away from the centre of the field of view.

To reduce the appearance of cone beam artefacts, multislice scanners with more than four slices utilise special reconstruction algorithms, which are either adaptations of conventional filtered back-projection methods or modified Feldkamp-type reconstructions.

Related Articles: Artefact, Beam hardening, Cone beam CT, CT reconstruction, Helical artefact, Image artefact, Metal artefact, Motion artefact, Partial volume effect (artefact), Ring artefact

Cone beam CT

(Diagnostic Radiology) Single-slice CT scanners were limited to a maximum coverage of 10 mm along the scan axis (z-axis). Because of the narrow divergence angle of the beam in the z-axis direction, they could be considered as having parallel ray geometry (Figure C.70a), with the x-ray source, detector and imaged slice lying in the same plane. Cone beam CT refers to the extended coverage of the x-ray beam along the z-axis, which is available on multislice scanners. The divergence angle of the beam is increased and this angle is referred to as the cone angle ($\alpha$) (Figure C.70b). For scanners acquiring more than four simultaneous slices, the beam can no longer be assumed to have parallel geometry and they are therefore considered as cone beam CT scanners. In cone beam CT, off-axis...
Cones and rods in vision

Consistency

The term consistency is used in quality control (QC) assessments. It describes the ability of a system to maintain stable parameters.

Confidence Limit

Confidence limits are outer limits making up a confidence interval and give a measure of the reliability of a single measurement based on statistics from a whole population. Confidence limits are set with respect to the expected probability level desired. The probability that a population contains a specific value is called the confidence level. In order to increase the confidence level, one must widen the confidence interval, i.e. change the confidence limit.

Conformal dose distribution

Conformal dose distribution is produced whenever each individual beam is shaped to the target volume as viewed from that beam position by the use of blocks, MLCs etc. The dose distribution isodose lines will then neatly surround the target with the appropriate level of dose while minimising the dose delivered to nearby healthy tissue and adjacent critical structures (organs at risk).

Abbreviation: MLC = Multileaf collimator.

Related Articles: Multileaf collimator, Tertiary collimator, Conformal radiotherapy, Custom blocking, Dose distribution, Target dose distribution, Critical structures, Organs at risk

Conformal radiotherapy

Conformal radiotherapy is used to describe any beam modulation that enhances conformation of the dose to the tumour. Three-dimensional (3D) conformal radiotherapy generally refers to multiple field treatments with each portal being reshaped with a multileaf collimator to account for the complex shape of the tumour. In this way, tumour is conformally treated and radiosensitive normal tissue is conformally avoided (see Figure C.72).

Figures C.73 and C.74 show a 3D view of a prostate treatment. In Figure C.73a, the patient outline can be seen (green transparent rendering), along with the prostate and seminal vesicles (target region) rendered in red. Three fields can be seen, each shaped with an MLC. In Figure C.74b, the high-dose region (95% of the isocentre dose) can be seen in white. Also seen are hot spots in the left and right lateral positions, often seen in this type of treatment.


Consistency

(General) See Retina

Confidence Limit

(Nuclear Medicine) Confidence limits are outer limits making up a confidence interval and give a measure of the reliability of a single measurement based on statistics from a whole population. Confidence limits are set with respect to the expected probability level desired. The probability that a population contains a specific value is called the confidence level. In order to increase the confidence level, one must widen the confidence interval, i.e. change the confidence limit.

Conformal dose distribution

(Radiotherapy) A conformal dose distribution is produced whenever each individual beam is shaped to the target volume as viewed from that beam position by the use of blocks, MLCs etc. The dose distribution isodose lines will then neatly surround the target with the appropriate level of dose while minimising the dose delivered to nearby healthy tissue and adjacent critical structures (organs at risk).

Abbreviation: MLC = Multileaf collimator.

Related Articles: Multileaf collimator, Tertiary collimator, Conformal radiotherapy, Custom blocking, Dose distribution, Target dose distribution, Critical structures, Organs at risk

Conformal radiotherapy

(Radiotherapy) When treatment is delivered with unshielded square or rectangular fields, the dose distribution mimics the beam arrangement, producing brick-shaped high-dose volumes. This result makes it difficult to avoid giving high doses of radiation to sensitive tissues around the treatment volume. This is because the high-dose volume does not have the same shape, i.e. does not conform to the shape of the intended treatment volume.

Conformal radiotherapy can be used to describe any beam modulation that enhances conformation of the dose to the tumour. Three-dimensional (3D) conformal radiotherapy generally refers to multiple field treatments with each portal being reshaped with a multileaf collimator to account for the complex shape of the tumour. In this way, tumour is conformally treated and radiosensitive normal tissue is conformally avoided (see Figure C.72).

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Consistency

(Diagnostic Radiology) The term consistency is used in quality control (QC) assessments. It describes the ability of a system to maintain stable parameters.
For example, x-ray tube output consistency refers to the ability of the x-ray system (the generator and the tube) to produce exposures with identical dose output (mGy) when all other parameters (kVp, mA, etc.) are constant. The minimal number of identical exposures (with identical parameters) used for calculating the consistency of an x-ray system is 4 (optimal is 6). The dose output consistency (%) in this case is calculated as the % ratio between the standard deviation of all four measured exposures (mGy) and the average of these (this has to be made for each focal spot) – 100*STDEV/AVERAGE. If the resultant figure is less than 5%, the dose output consistency is acceptable.

A similar method and calculation can be made to calculate, for example the consistency of the kVp (again the lesser the % the better the consistency).

The consistency described earlier is a widely used practical term. However, it can be misleading as it actually represents the error of consistency (1% consistency describes an excellent x-ray system, but actually it means that 99% of all exposures will be consistent).

The methods to perform various QC measurements are described in detail in the EMERALD materials.  
Related Articles: Accuracy, Precision 
Hyperlink: EMERALD: www.emerald2.eu

**Constructive interference**  
(Ultrasound) Two waves that travel together can, dependent on their respective phase, be observed as a wave that is the sum of the two waves’ individual amplitudes (constructive interference), or add up to no apparent wave motion at all if the amplitudes are equal and the phases are opposite (destructive interference). In Figure C.75, two point sources emit continuous waves, and in certain directions the waves appear to be in phase, whereas in others, to be out of phase. As can be deduced from the figure, this depends on the difference in distance from the observation point to the respective sources. Distance differences that correspond to an integer number of wavelengths result in constructive interference.

**Constructive interference steady state (CISS)**  
(Magnetic Resonance) The CISS pulse sequence was constructed to reduce the banding artefacts that might appear in balanced SSFP pulse sequences (true-FISP). The idea is to collect two true-FISP pulse sequences where one of them is acquired with a phase cycling scheme of the RF pulse, i.e. 20°, -20°, -20°, -20°, ..., and the other is not, i.e. 20°, -20°, -20°, -20°...

In Figure C.76, a motion compensating, fully refocused 3D SSFP-fid image of the internal auditory canal and the inner ear in a patient with normal findings is shown.

(See colour insert.) Three-field treatment of a prostate in red showing the fields and the MLC shapes for each beam (VARIAN CadPlan planning system).

(See colour insert.) The white area represents the high dose envelope (95% of the iso-centre/prescribed dose) (VARIAN CadPlan planning system).
In modern brachytherapy dosimetry, reference air kerma rate or air kerma strength is the quantity used to calculate absorbed dose. See Source strength for a full description of specification of source strength.

**Related Articles:** Source strength, Mass of radium, Equivalent mass of radium, Apparent activity, Reference air kerma rate, Air kerma strength

**Contamination**

(Nuclear Medicine) In nuclear medicine, a radioactive contamination can refer to an unintentional or accidental spill of radioactive material on to any material, surface, environment or individual during a working operation, for example labelling, injection, taking of specimens or blood samples. In the specific case of the human body, this radioactive contamination includes both external skin contamination and internal contamination, irrespective of route of intake.

If a radionuclide used in nuclear medicine is accidentally spilled, the material could be spread and spoil measurements of other radioactive samples or worse constitute an undesired source of exposure. A surface contamination is expressed in units of activity per unit of area, Bq/cm². The national legislation normally states the maximum permissible levels of exposure and contamination, i.e. the amount of activity per unit area depending on the radiotoxicity of the radionuclide.

When checking a workbench in a laboratory using radionuclides, a simple hand-detector giving the count rate may be adequate. However, if the radionuclide is unknown, a more accurate assay is needed, for example a gamma spectroscopic analysis. Monitors showing the activity per unit of surface (Bq cm²) are available from various companies.

An internal contamination denotes that a radioactive contamination has entered the body through ingestion or inhalation. This will contribute to radiation dose of the individual. It is very important to use personal protective equipment, such as gloves and a white coat when working with radiopharmaceuticals. After a completed operation, the working bench should be monitored with respect to surface contamination. For the measurement of a whole body content of activity, counting in a special whole body counter may be necessary. Note that if a whole body counter is not available, a scintillation camera without the collimator may be used in special cases (depending on the background).


**Contamination monitoring**

(Radiation Protection) Contamination monitoring seeks to determine if radioactive contamination has occurred, and if so, where and how much. The contamination may be to a person, or to a surface (wall, floor, furniture, etc.). Contamination monitoring is carried out using an appropriate monitor, sensitive to the radioisotopes in use in the area. For contamination involving gamma-emitting radionuclides, a scintillating crystal–photomultiplier detector is required. For contamination involving medium- to high-energy beta-emitting radionuclides, Geiger–Müller detector is required. While for low-energy beta and alpha particles, contamination monitoring must be carried out using a wipe test.

Contamination monitoring should be carried out after any work with unsealed sources, and the results recorded.

**Related Article:** Wipe test

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**FIGURE C.76** A motion compensating, fully refocused 3D SSFP-fid image of the internal auditory canal with normal findings.

**Related Articles:** Fast imaging with steady-state precession (FISP), SSFP, Steady-state free precession

**Contact therapy**

(Radiotherapy) Contact therapy describes treatment with low-energy kilovoltage x-ray beams (accelerating potential 40–50kV). The SSD for these treatment fields is typically a few centimetres or less. Contact therapy is used for superficial treatment depths of around 2 mm.

**Abbreviation:** SSD = Source to skin distance.

**Related Article:** Superficial therapy

**Contained activity**

(Radiotherapy, Brachytherapy) Calibration of source strength is a very important part of a comprehensive brachytherapy quality system. The instruments, ion chambers and electrometers, used for source strength determinations, should have calibrations that are traceable to national and international standards.

**Specification of Source Strength for Photon-Emitting Sources:** Source strength for a photon-emitting source can be given as a quantity describing the radioactivity contained in the source or as a quantity describing the output of the source:

1. Specification of contained activity
   a. Mass of radium; mg Ra
   b. Contained activity; Ci, Bq
2. Specification of output
   a. Equivalent mass of radium; mg Ra eq
   b. Apparent activity
   c. Reference exposure rate
   d. Reference air kerma rate
   e. Air kerma strength

Contained activity is a quantity that can be used for all types of brachytherapy sources.

For brachytherapy dosimetry, the quantity of interest is the output of the encapsulated source, not the contained activity. (Sources are encapsulated, and it is thus difficult to determine the contained activity.) The quantity apparent activity, which is an output specification, has been used as an alternative to contained activity and it is still used, especially for radiation protection applications.
Contingency plan

(Radiation Protection) Whenever ionising radiation is used in applications that can create hazard for the workers and the population, it is essential to prepare contingency plans in order to minimise the possible risk and provide a prompt response in case of incidents/accidents.

In general, contingency plans mean comprehensive systems of emergency response capabilities, which involves the national authorities and promote overall coordination among the hierarchy of emergency response organisations at regional and local level. Therefore, depending on the specific applications, contingency plans might include radiation protection measurements for people and environment, handling of health effects, managing of radioactive material, managing of radioactive waste, cleaning up of sites etc.

Although there are unfortunate examples of major accidents related to medical radiation sources, usually, in medical applications, contingency plans should be handled at local and regional level.

Starting from the basic, there must be a clear identification of the major principal causes of incidents/accidents, with clear identification of roles and competences. Contingency plans include the detection of the event, the involvement of qualified staff, the support of laboratory services and eventually other structures.

In medical applications, priority should be given to the safety of the patients.

Contingency plans shall be studied in detail by the radiation protection specialists and tailored to each equipment/structure, depending on the kind of possible risk.

Hyperlink: IAEA: http://www.iaea.org

Continuous slowing down approximation

(Radiotherapy) The various types of interactions, which a charged particle experiences during its passage through the medium, produce a gradual loss of energy, bringing the particle eventually to stop. The track length distributions and dose deposition of a charged particle interacting with matter depend on the approximation employed in defining the transport process. A common approximation in use is the continuous slowing down approximation (CSDA). In CSDA, the charged particles are assumed to lose their energy continuously at a rate that is given by the stopping power \( S(T) \), where \( T \) is the particle kinetic energy. In CSDA, it is supposed that no knock-down electrons (\( \delta \) particles) or Bremsstrahlung photons are created. The kinetic energy lost is locally converted into imparted energy. The track length distribution \( y(T) \) takes the following form:

\[
y(T)dT = \frac{dT}{S(T)}
\]

When the fluences \( \Phi \) are calculated using the CSDA approximation, the absorbed dose is obtained by

\[
D = \int_{0}^{\infty} \Phi \frac{S(T)}{\rho} dT
\]

The CSDA inadequately describes the electron transport. In the extreme case of a head-on collision between electrons, all of the incident energy will be given to the knock-on electron. Because the lack of identity of electrons, it is usual to identify the electron with lower energy after the collision as secondary electron. An electron may therefore lose as much as half of its kinetic energy in a single atomic collision and the statistical fluctuations in the energy losses are significant. The generation and transport of high-energy secondary electrons (\( \delta \) electrons) was taken into account by the Spencer-Fano model (Spencer and Fano, 1954). An energy limit \( D \) is chosen in this model. Energy losses above \( D \) are considered to result in the production of \( \delta \) electrons while those below represent the imparted energy to matter. The initial kinetic energy of the \( \delta \) electron equals the energy lost by the primary electron, so no energy is imparted to matter in this process.

Abbreviation: CSDA = Continuous slowing down approximation.

Related Articles: Mass collision stopping power, Mass stopping power, Electron stopping power


Continuous spectrum

(Diagnostic Radiology) A continuous spectrum is one in which there are photons at all energies within the range of the spectrum. The Bremsstrahlung process produces a continuous x-ray spectrum (also called white spectrum).

Related Article: Bremsstrahlung

Contralateral

(General) Directional anatomical terms describe the relationship of structures relative to other structures or locations in the body.

‘Contralateral’ means on different sides of the midline. The right shoulder and left hip are contralateral to each other.

Related Article: Anatomical relationships

Contrast

(General) For the human eye, contrast is a measure of the ability to distinguish between two adjacent objects or an object and the background. In a digital image, the contrast is defined by the signal difference between the two objects. A common goal for all disciplines in medical imaging is to increase the contrast between an object of interest and the surrounding tissue so that any pathological tissue or bioprocess becomes evident to the radiologist. In diagnostic radiology and MR, one uses contrast agents (or contrast media) to increase the difference in acquired signal.

Contrast agent

(Magnetic Resonance) In medical imaging, a contrast agent is an exogenous substance introduced into the body by some means (e.g. by injection or orally) in order to modify the signal from certain tissues or structures, hence making them more conspicuous in the acquired image. Contrast agents are therefore chosen for their ability to modify a property of the tissue that has an impact on its appearance in the image, as well as for their suitability for use in vivo in terms of safety and practicality.

In x-ray imaging, the most commonly used intravenous contrast agents are iodine based. Barium-based contrast agents are usually used orally (e.g. barium meal used for x-ray examination of the hollow gastrointestinal tract). The high subject contrast in these cases requires use of higher energies (kVp).

In MRI, the ability to modify image contrast by changing acquisition parameters and by tailoring the pulse sequence to produce different weightings arguably reduces the need for exogenous agents as compared, for example to x-ray imaging. However, there are a number of situations in which MRI contrast agents are useful, and their use has become widespread.
Applications of contrast agents in MRI include the following:

- Enhancement of signal from selected tissues or areas of pathology, usually on the basis of vascularity
- Enhancement of signal from flowing blood in contrast-enhanced MR angiography
- Elimination of signal from organs that might otherwise obscure structures of interest or cause artefacts, such as the bowel
- Dynamic studies of perfusion or function, sometimes for tissue characterisation
- In emerging molecular imaging techniques

To be useful as a contrast agent for MRI, a material must have some effect on one or more of the NMR properties of tissues – usually the $T_1$ and/or $T_2$ relaxation times. The most commonly used relaxation agents are metal-chelate complexes containing lanthanide ions, predominantly gadolinium. There are also agents based on small particles of paramagnetic iron oxide, which operate primarily as $T_2$-shortening agents. These particles may be bound to receptor-specific carbohydrates or to antibody fragments for enhanced specificity, for example in hepatic or lymph node imaging.

These early examples of targeted agents are leading to the development of MRI molecular imaging agents, such as $\alpha$,$\beta$,-targeted paramagnetic nanoparticles to image angiogenesis. Such studies may make use of nuclei that are not present in the body in significant quantities, such as fluorine, blurring the distinction between contrast agents and tracers.

Oral contrast agents are designed to manipulate signal from the bowel, either to enhance it (gadolinium agents, ferric ammonium citrate or oil emulsions) or, more commonly, to suppress it (iron oxide beads, carbon dioxide, barium sulphate or perfluorochemicals).

For vascular imaging, conventional gadolinium agents are frequently used, but are limited by their relatively rapid extravasation. A number of longer lasting blood pool agents have been developed, and such agents based on gadolinium are now entering clinical use.

Gadolinium-based agents are normally used in conjunction with $T_1$-weighted imaging, with areas of high uptake appearing bright due to $T_1$ shortening. Iron oxide–based agents, on the other hand, produce dark regions on $T_1$-weighted images due to $T_2$ shortening. These are known colloquially as ‘hot’ and ‘cold’ spots, respectively.

Related Articles: Gadolinium, Negative contrast media, Paramagnetic contrast agents, Positive contrast media, USPIO (ultra small particles of iron oxide)

Contrast agent
(Ultrasound)

Background: Contrast agents for ultrasound are a fairly new aid to improve the diagnosis, used in cardiology, and for identification and classification of tumours, especially in the liver. Essentially, the contrast agents are microbubbles with a size that is less than the capillary diameter (i.e. a mean size of ~2μm). The microbubbles need to be stabilised by a shell, which usually is composed of lipids or proteins. The gas is also preferably one that has low diffusivity, such as perfluorocarbons or sulphurhexafluoride.

Acoustic Properties: A free gas bubble in a fluid acts as a mass-and-spring resonance system, where the gas pressure acts as the spring and the displaced fluid the mass. Consequently, the bubble has a resonance frequency given by

$$\omega_{res} = \frac{1}{R_0} \sqrt{\frac{3\gamma p}{\rho}} \tag{C.5}$$

where
- $\omega_{res}$ is the resonance frequency
- $R_0$ is the equilibrium radius of the bubble
- $\gamma$ is the ratio of specific heat of a gas at constant pressure and at constant volume
- $p$ is the hydrostatic pressure outside the bubble
- $\rho$ is the density of the surrounding fluid (Leighton, 1994)

Consequently, a 2μm air bubble in blood will resonate at 3.26 MHz, in perfect range for diagnostic ultrasound. This dramatically increases the detectability of small gas bubbles with ultrasound. The drawback is that so small bubbles dissolve in a matter of milliseconds, due to surface tension, which is why contrast bubbles need to be stabilised by a shell.

Detection Principles: An interesting property of bubbles is that they behave strongly non-linear to acoustic stimuli, already at relatively small acoustic pressures. The basic underlying mechanism is that the bubble tends to expand more easily than being compressed. Thus, harmonics of the incident ultrasound wave are easily produced. One of the first approaches was to simply transmit at one frequency, but have the detection at twice that frequency (harmonic imaging). This assumes a transducer that is wideband enough, and also that the pulses need to be fairly narrowband in order to avoid spectral leakage from the fundamental to the harmonic. In order to increase the bandwidth of the pulse, and thereby the spatial resolution, a detection scheme was developed where two pulses of opposite polarity are transmitted (commonly referred to as pulse inversion). For a linear scatterer, the second received signal will be identical to the first but inverted. On the other hand, a non-linear scattering object will produce signals that are not similar. When adding the two received signals, linear echoes will cancel while non-linear give rise to a detectable signal. Several variants of this scheme are now in use, notably one where the pulses still have the same phase, but different amplitude. As a general rule, detections schemes are then divided into single-pulse (such as harmonic imaging) and multi-pulse (such as pulse inversion) methods.

Related Articles: Contrast media, Echo-enhancing agent, Microbubbles


Contrast detail
(Diagnostic Radiology) The visibility of an object in an x-ray image depends on both its physical contrast and its size (detail). The reduction in visibility relating to an object’s physical contrast is caused by limitations of the contrast sensitivity of the imaging process and by the presence of visual noise. The reduction of visibility relating to object size (detail) is because of the blurring that occurs during the imaging process. The visibility, especially for small objects that also have relatively low physical contrast, because they are thin in the direction of the x-ray beam, is affected by a combination of the three factors; contrast sensitivity, blurring and noise. Test devices that contain objects that have a combination of different physical contrasts (object thickness) and detail (size) are used to evaluate imaging procedures. An image of the device is produced and it is then determined which objects are visible.

The contrast detail diagram is a graph or curve plotting the ‘just visible’ objects using (physical) contrast versus detail (size) scales. It is a qualitative method combining both the concepts of contrast resolution and the spatial resolution. The assessment (observation) aims to detect the minimal contrast necessary to visualise an object with certain size. These minimal (limiting) contrasts are then presented against the object size, thus forming a diagram with contrast on the y-axis and detail size on the x-axis. The diagram divides the
space into two parts – visible (above the curve) and invisible (under the curve). If the curve is closer to the x-y axes, this indicates better contrast detail for certain system (larger visible part). Also, the curve shows that small details require high contrast in order to be visible.

Figure C.77 shows a test object used to examine contrast detail of an x-ray system (usually fluoroscopic system). The test object is a Leeds TO10. It includes a number of rows, each with identical detail sizes. Within each row, the contrast gradually decreases (by gradually decreasing the absorption of the objects). The observer has to identify the object in a certain row, where its image cannot be distinguished anymore from the background (the limiting contrast). The value of this contrast (taken from a table accompanying the test object) is then plotted against the diameter of the object.

Figure C.78 shows typical contrast detail diagrams of three image intensifiers (II): old II with 25 cm diameter; same old II with 15 cm diameter (zoomed image); a new II with 17 cm diameter. It is obvious that the third curve is closer to the x-y axes and represents a system with better imaging characteristics. The diagram shows that using the third II (17 cm), one needs 0.35 (35%) contrast in order to see a detail with 0.5 mm size. Respectively, if the first II (25 cm) is used, such detail can be seen only if its contrast is 0.67 (67%).

Contrast detail and contrast resolution concepts can be described also with the ratio

$$\frac{(C \ast D)}{N} = \text{const}$$

where

- $C$ is the contrast
- $D$ is the detail diameter
- $N$ is the noise level

**Related Articles:** Contrast resolution, Spatial resolution

**Hyperlink:** EMERALD: [www.emerald2.eu](http://www.emerald2.eu)

**Further Reading:** Oppelt, A. ed. 2005. Imaging Systems for Medical Diagnostics, Siemens, Erlangen, Germany.

**Contrast detail (C-D) studies** *(Diagnostic Radiology)* See Contrast detail

**Contrast-enhanced angiography** *(Magnetic Resonance)* Blood vessels can be visualised with MR in several ways (magnetic resonance angiography, MRA). One method uses contrast-enhancing agents (contrast-enhanced MRA, CE-MRA). The contrast agents involved are paramagnetic, and affect the magnetic properties of the tissue they come in contact with. When injected into the blood stream, the contrast media reduces the $T_1$ of blood, and CE-MRA exploits this difference in $T_1$ between tissue and blood to visualise the blood vessels while suppressing signal from the background tissue.

A CE-MRA sequence is most often a 3D gradient-echo sequence with a short $TR$, to suppress the signal from background tissue with its relatively long $T_1$. A short $TE$ is used to minimize $T_2^*$ effects due to the contrast agent. As the high signal in the vessels is primarily due to the presence of contrast media, blood with low velocity is enhanced, as well as fast flowing blood. This is an advantage when investigating veins, or aneurysms and other pathologies. The data are very often post-processed with maximum intensity projection (MIP) to obtain a 3D overview of the vessel tree.

As there is a time delay between the injection of contrast agent and the arrival of this agent to the region of interest, a timing method has to be used. It can consist of a test bolus, which is timed, and the measured delay time entered into the CE-MRA sequence. It can also be in the form of a real-time view of the area of interest, where the CE-MRA is interactively started as the contrast reaches the vessels to be studied.

With fast sequences, it is possible to retrieve data for different time delays, as is seen in Figure C.79. The second image is taken at a later time than the first one.

CE-MRA is clinically used for a variety of applications, for example to examine vessels in the head–neck, thorax, abdomen or extremities.

**Contrast enhancement** *(Diagnostic Radiology)* Contrast enhancement is a process that is used to increase the contrast and visibility of specific organs, vessels or tissues in medical imaging. A variety of contrast agents are used to enhance the physical contrast within the body.

X-ray spectrum optimisation and techniques such as digital subtraction angiography (DSA) are used to enhance contrast during the acquisition of x-ray images.

Digital processing is used to enhance and optimise contrast, especially in digital radiography.
Contrast improvement factor (CIF)

(Diagnostic Radiology) The contrast improvement factor (CIF) is the ratio of the contrast when a specific grid is used compared with the contrast without the grid. It is a function of the grid characteristics and the amount of scattered radiation from the patient’s body. As an example, CIF values can be 2.4, 3.3 etc.

CIF is one of the measures of efficiency of anti-scatter grids. The other methods to measure this are the Bucky factor (the ratio of the radiation entering the grid to the radiation passing through the grid) and the primary transmission factor (the percentage of primary radiation passing through the grid).

Contrast media

(Nuclear Medicine) In medical settings, a contrast media (or contrast agent) is administered to achieve a higher contrast difference between a structure or blood flow and the surrounding matter. In conventional x-ray imaging, contrast media can be used to enhance the visibility of blood flow, for example in ventilation and perfusion lung studies or in angiographic investigations.

A contrast media with high atomic number relative to the surrounding matter increases the photon attenuation; hence, regions with high accumulation of contrast media will appear darker compared to an identical situation without the contrast media. As an example, iodine-based contrast media is used for x-ray angiographic examinations.

Gadolinium (Gd) is used as a contrast media in MR because the presence of Gd molecules induces a quicker relaxation in adjacent water molecules.

Even though the examples are few, the contrast media has induced allergic reactions in some patients and the use of such should therefore be under supervision.

Contrast media

(Ultrasound) Micorobubbles are used as contrast media in ultrasound examinations. These are micron-sized bubbles, stabilised by a shell. Further details are found under contrast agents.

Contrast resolution

(Diagnostic Radiology) Contrast resolution is a characteristic of an imaging process relating to the ability to see, or resolve, small differences in contrast (low contrast) of objects or areas within an image. The minimal (limiting) contrast to be seen in one imaging system is highly related to the maximal noise in this system. This way contrast resolution can be quantitatively described through the signal to noise ratio (SNR) or noise power spectrum (NPS) of the system. This SNR measurement is especially suitable for digital imaging systems.

A simple presentation of contrast resolution is the minimal (limiting) contrast, which the system can discriminate. For example, a computed tomography (CT) system discriminates contrast (density difference) \( \sim 0.25\% \), while a classical x-ray radiography system discriminates \( \sim 10\% \) contrast.

Various imaging systems have specific ways to measure their intrinsic contrast resolution – for example, CT densitometry can be used in CT; Pulse overlap separation can be used in ultrasound; contrast detail diagram can be used in x-ray fluoroscopy (qualitative measurement), etc.

Contrast resolution (and contrast detail) concepts can be described also with the ratio

\[
\frac{(C \times D)}{N} = \text{const}
\]

where

- \( C \) is the contrast resolution
- \( D \) is the detail diameter
- \( N \) is the noise level

This constant is maintained for a wide range of diameters and is independent of the spatial resolution of the imaging system.

Related Articles: Contrast detail, Signal to noise ratio, Noise power spectrum, Spatial resolution


Contrast resolution

(Ultrasound) In medical imaging generally, the term contrast resolution describes the ability of a modality to display differences in...
tissues resulting from the different properties of those tissues. The ability to discriminate between different tissues is key to the performance of the imaging system.

For ultrasound, changes in acoustic properties at a boundary cause reflection from that boundary. The brightness is dependent on the difference in acoustic impedance, and size relative to the wavelength and orientation relative to the beam. Within tissue, the characteristic brightness is dependent on the scattering within the tissue. The perception of differences in acoustic properties in an ultrasound scan is complex and is dependent on, for example

- The magnitude of differences
- The instrument settings used (e.g. power, gain, dynamic range)
- The size of the target
- The presence of artefacts (e.g. speckle)
- Whether defined edges are present
- Viewing conditions

Tests of contrast resolution are difficult for the same reasons. Figure C.80 shows an ultrasound phantom, which contains large targets of different scattering levels. The same phantom has high-contrast cyst-like targets, which become difficult to visualise when small, illustrating the influence of target size on contrast.

On Figure C.80, the +3dB and −3dB targets are difficult to visualise and the fine speckle pattern contains significantly higher variation, which detracts from the larger target. The high-contrast cyst-like targets (white box) of 4, 3 and 2 mm diameter are not seen throughout the image; the 2 mm targets are not imaged at depth due to poor spatial resolution and speckle.

**Related Articles:** Spatial resolution, Phantom, Speckle, Dynamic range

**Contrast scale**
(Diagnostic Radiology) The contrast scale is a term related to the subject contrast in x-ray imaging (but also in any other imaging). The contrast scale of a black–white x-ray image refers to the number of visibly distinguishable shades of grey in the x-rayed area.

Two terms often used in practice are long-scale contrast and short-scale contrast. Usually, long-scale contrast can be referred to a film with wide latitude. Such contrast can also be called low contrast or soft contrast. Similarly short-scale contrast can be referred to a film with narrow latitude. Such contrast can also be called high contrast or hard contrast.

These qualitative practical terms can often be confusing. In principle, the contrast scale and the degree of image contrast depends on the absorption of x-rays by the individual structures/ organs. This way, it is a function of the x-ray spectrum, the organs attenuation, the x-ray film characteristics etc.

In digital imaging, the contrast scale directly depends on the parameters of the window width (WW). Large window produces long-scale contrast or soft contrast, while narrow window produces short-scale contrast or hard contrast.

For illustrations to the preceding text, see the images in the article, High contrast.

**Related Articles:** Subject contrast, High contrast, Latitude, Window


**Contrast scale**
(Ultrasound) A contrast scale shows a range of grey levels from white to black typically divided into 16 discrete levels (Figure C.81). The level of grey can be described as a percentage, for example 100% as white, 0% as black or vice versa for a greyscale image printed on white paper.

Contrast scales are used as an aid in setting up monitor brightness and contrast so that the full range of greys in the processed image can be displayed for optimum contrast resolution.

**Related Article:** Contrast resolution

**Contrast sensitivity**
(Diagnostic Radiology) Contrast sensitivity is a characteristic of imaging systems, including the human visual system, which determines the lowest contrast objects that are visible (Figure C.82).

The contrast sensitivity of a specific imaging procedure is determined by the physical principles of the process (CT and radiography are very different) and the operating factors. The same contrast sensitivity is not required for all procedures and should be adjusted to optimise visibility for the specific clinical need.

**Contrast threshold**
(General) Contrast threshold is the smallest contrast in luminance/brightness that is perceptible to the human eye. It is dependent on...
the angular size of the target and varies between observers and over time in an individual. The contrast threshold also depends on the expectation and probability that a target will appear.

**Related Articles:** Contrast scale, Perception

### Contrast to noise ratio

*(Nuclear Medicine)* The contrast to noise ratio (abbreviated CNR) is a quantification of a medical imaging modality system’s ability to distinguish between structures and noise in an acquired image. CNR can be quantified with equation C.6:

\[
\text{CNR} = \frac{(S_A - S_B)}{\text{Noise}} \quad (C.6)
\]

where \(S_A\) and \(S_B\) are the average signal strength in tissue \(A\) and \(B\), respectively. The noise is usually measured as the standard deviation in a region of interest (ROI) outside any biological structure.

### Contrast transfer function

*(Diagnostic Radiology)* The contrast transfer function (CTF) is used in diagnostic radiology quality control (QC). It represents the change of contrast as a function of the spatial resolution (contrast % decrease related to increased lp/mm). CTF presents results comparable with those of the modulation transfer function (MTF). These results lack the precision of the MTF but are more easily obtained. MTF has a theoretical base and shows precise results, but requires special hardware and software.

A typical example of CTF used in practice will be from QC of a computed tomography (CT) system. Figure C.83 shows a bar pattern test phantom with known contrast values (CT numbers, CT#, or HU) of the materials (in the case Perspex with CT# +120 HU and water with CT# 0 HU, presenting 100% contrast as +120 CT numbers). The density profile through the CT scan of the phantom will show the maximal resulting contrast of each bar pattern (e.g. 108 CT# for the bar pattern with 3.1 lp cm – the first on the right). This way, plotting the % of contrast decrease against the spatial resolution of each bar pattern will present the CTF. Figure C.84 presents three CTF functions made with different scanning parameters.

CTF can present indicative practical information about the relative change of the image quality parameters of an imaging system.

**Related Article:** MTF

### Control button

*(General)* A control button (also called as push-button, pushbutton or simply button) is a switch mechanism for controlling operation of the electrical equipment (e.g. x-ray equipment). The push-to-make switch contains a spring, returning the actuator to a certain position. Contact is made when the button is pressed and is broken when the button is released. The exposure control of x-ray system terminates the x-ray exposure when pressure is released from the exposure control button.

An x-ray unit may comprise a two-step control button formed of a standby button (e.g. for starting rotation of the anode) and an execution (exposure) button.

An example of a control button of mobile x-ray equipment is shown in Figure C.85.

**Related Articles:** Switch, Dead man’s switch, Exposure switch, Foot switch

### Control panel

*(General)* See Control button
Control rods of nuclear reactor

(Radiation Protection) Nuclear reactors are based on the fission reaction \((n, f)\), which occurs for heavy nuclei (mass number \(A > 220\), e.g. \(^{235}\text{U}, ^{239}\text{Pu}, ^{233}\text{U}\)) induced by neutrons. An example of this nuclear reaction induced by the thermal energy neutron (0.025 eV) is

\[
^{235}\text{U} + \frac{1}{2}n \to ^{236}\text{U}^* \to ^{99}_{45}\text{Mo} + ^{135}_{50}\text{Sn} + 2\frac{1}{2}n
\]

+ ~ 200 MeV energy released

Many other nuclides besides those mentioned earlier are also produced.

The energy produced in nuclear fission by 1 kg of uranium is about \(10^{14}\) J (in burning 1 kg of coal is \(3 \times 10^7\) J). The daily fuel requirements for 1 GWe power plant are 3 kg of uranium and 8 \times \(10^6\) kg of coal.

The neutrons produced in fission (usually 2 or 3) may evoke another fission and thus initiate the self-sustaining chain reaction if the fissionable fuel amount is equal or greater than its critical mass. The critical mass of the fissile material may be controlled by absorbing neutrons. Cadmium absorbs neutrons very efficiently by its extremely large cross section.

The most important components for a controlled nuclear reactor are (1) fissionable material (located in fuel rods), (2) moderator to slow down neutrons, (3) control rods and (4) radiation shield. In Figure C.86, the schematic cross section of the reactor core is shown. The role of the cadmium control rods is to maintain a self-sustained chain reaction or to break it off. It is realised by removing or positioning cadmium rods in the reactor core.


Controlled area

(Radiation Protection) **Classification of Areas**: The responsibility of designating the controlled areas is attributed to the registrants and licensees, who may appoint qualified experts to deal with the practical work. They shall designate as a controlled area, any area in which specific protective measures or safety provisions are or could be required for (1) controlling normal exposures during normal working conditions and (2) preventing or limiting the extent of potential exposures.

Convection

(Radiation Protection) **Convection**: Convection refers to the transport process of molecules in gases and liquids and it is one of the major methods for heat and particle transfer. Convective heat and mass transfer in liquids are the sum of two separate processes: diffusion and advection. Diffusion refers to the rearrangement of particles due to their individual Brownian motion and advection to the large-scale currents in the fluid. The converse of convective heat transfer is conductive heat transfer where the energy propagates through the fluid via vibration interactions between individual atoms or molecules.
Converging collimator

(Nuclear Medicine) The holes in a converging collimator converge to a point in front of the collimator, typically at a distance of 40–50 cm. With this type of collimator, it is possible to attain magnified images. The collimator projects a magnified image if the object is placed between the front of the collimator and the point of convergence. An object placed further away will be imaged mini-


dified and inverted, but these collimators are rarely used for such purposes. The magnification factor is the relationship between the image size \( I \) and object size \( O \), as given by

\[
l = \frac{I}{O} = \frac{(f + t)}{f + t - b}
\]

where

- \( f \) is the collimator front to convergence point distance
- \( b \) is the collimator to source distance
- \( t \) is the collimator thickness

Since the magnification depends on the collimator to source distance, sources at different depths have a different magnification factor. Differences in object depth, i.e. different magnifications, may result in image distortions and the user must be well aware of this effect or else it may ultimately lead to misdiagnosis and/or ineffective/harmful treatment.

The converging collimator contribution to the system resolution is

\[
R_\text{coll} = \left[ \frac{d[l'(f + t) + b]}{l'_\text{eff}} \right] \times \left[ \frac{1}{\cos \theta} \right] \times \left[ 1 - \left( \frac{l'_\text{eff} / 2}{f + l'_\text{eff}} \right) \right]
\]

where

- \( d \) is the hole diameter
- \( l'_\text{eff} \) is the effective diameter, accounting for septal penetration
- \( \theta \) is the angle between the central axis of the collimator and an off-center source as seen in Figure C.87

From this equation, it is possible to see that the spatial resolution is best along the central axis.

The geometric efficiency of a converging collimator can be expressed as follows:

\[
g = K^2 \left( \frac{d}{l'_\text{eff}} \right)^2 \times \left[ \frac{d^2}{(d + t)^2} \right] \times \left[ \frac{f^2}{(f - b)^2} \right]
\]

where \( K \) is a constant depending on hole shape. The maximum efficiency is obtained at the point of convergence. At typical imaging distances, 5–10 cm, the converging collimator offers a good combination of both spatial resolution and efficiency. The projected image is magnified, which means that the field of view must be larger than the object itself and as a result, imaging with converging collimators is best suited for large-area detectors.

Related Articles: SPECT, Parallel-hole collimator, Diverging collimator, Collimator design, Collimator parameters


Conversion efficiency of photocathodes

(Nuclear Medicine) The conversion efficiency or quantum efficiency of a photocathode is the quotient between the number of photoelectrons emitted and the number of incident photons. Photocathode efficiency is on the order of 20%–30%. The photocathode thickness must be minimised to allow the excited electrons to reach the surface with enough energy to penetrate the surface potential barrier. The threshold depth at which electrons can originate and reach the surface and penetrate the surface potential barrier is called the escape depth. For metals, the escape depth is a few nanometers and for some semiconductors it can extend to about 25 nm. This layer is not thick enough to attenuate all the incoming light photons.


Converter

(General) In electronics, a converter is a device (e.g. rectifier, inverter) that transforms one or more parameters (e.g. frequency, voltage, number of phases) of electrical power from one value to another. An analogue-to-digital converter (abbreviated ADC, A/D or A to D) is a device that converts continuous signals to discrete (digital) values. The reverse operation is performed by a digital-to-analogue converter (DAC).

Related Articles: Analogue-to-digital converter (ADC), Digital-to-analogue converter (DAC), Frequency converter, Ramp converter, Scan converter, Digital scan converter, Wilkinson converter

Convex array

(Ultrasound) See Curvilinear array transducer

Convex target volume

(Radiotherapy) A convex target volume is defined as one whose interior angles are all less than or equal to 180°. The opposite of a convex target volume is a concave target volume, see Figure C.64 for theoretical examples.

Convex target volumes can be treated successfully with multiple beams, for example each beam may conform to the shape of the tumour presented – the Beam’s eye view. See Conformal radiotherapy for more information.

However, if critical organs are close to the tumour site, then the target volume will often be defined as concave, with the aim to limit the dose distribution within the critical organ. Typical tumour
sites and respective critical structures that require concave target volumes include
- Pharynx (spinal cord)
- Prostate (rectum)
- Skull base (optic pathway structures, brainstem)

In order to achieve a concave dose distribution to match the concave target volume, techniques such as intensity modulated radiotherapy are required.

Related Articles: Target volume, Conformal radiotherapy, Beam’s eye view, Intensity modulated radiotherapy


**Convolution**

(Convolution) Convolution is a mathematical operator that operates on two functions to produce a third function. The final function is typically a modified version of one of the two original functions, for example in image processing where one function represents the image and the other function a filter (e.g. smoothing and edge enhancing). The resulting function is then a smoothed version of the initial function, giving a smoothed image with increased signal to noise ratio (SNR).

Related Articles: Signal to noise ratio, SNR, Spatial filtering, Kernel

**Convolution integral**

(General) Convolution is the basis of many data and image processing techniques. It is a mathematical process similar to cross-correlation for real data:

\[(f \ast g)(t) = \int f(\tau)g(t-\tau)d\tau\]

As can be seen from the convolution integral, the convolution of two functions is defined as the integral of the product of f with a reversed and shifted g. It is denoted with the star symbol.*

It is used to describe the performance of linear time invariant (LTI) systems, which are characterised by the impulse response h(t). For example, if the input is x(t), then the output will be

\[y(t) = (x \ast h)(t)\]

Convolution can be carried out in single or multiple dimensions. In one dimension (i.e. time), the process has the effect of filtering the data in some way. The convolution of a 2D image with a 2D mask can provide many practical image processing procedures such as sharpening, edge detection etc.

**Convolution Theorem:** The Fourier transform of a convolution equals the multiplication of the individual Fourier transforms:

\[\mathcal{F}(f \ast g) = \mathcal{F}(f) \mathcal{F}(g)\]

**Discrete Convolution:** The convolution of discrete functions is as follows:

\[(f \ast g)(n) = \sum_{m=-\infty}^{\infty} f[m] \cdot g[n-m]\]

**Convolution Identities:**

<table>
<thead>
<tr>
<th>Property</th>
<th>Equation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Commutativity</td>
<td>( f \ast g = g \ast f )</td>
</tr>
<tr>
<td>Associativity</td>
<td>( f \ast (g \ast h) = (f \ast g) \ast h )</td>
</tr>
<tr>
<td>Distributivity</td>
<td>( f \ast (g + h) = (f \ast g) + (f \ast h) )</td>
</tr>
<tr>
<td>Identity</td>
<td>( f \ast \delta = \delta \ast f = f )</td>
</tr>
</tbody>
</table>

**Related Article:** Fourier transform

**Convolution kernel**

(General) See *Convolution method*

**Convolution method**

(General) The convolution method is a model-based dose calculation model, which directly computes the dose to a patient. The dose calculation explicitly takes into account the energy and the geometry of the beam, the presence of beam modifiers and also the patient representation, which is usually obtained from image CT data.

A model-based method computes the dose per unit fluence or energy fluence, allowing expression of the beam monitor units in a phantom-independent manner such as energy fluence per monitor unit.

The convolution method separates the primary photon interactions from the transport of any secondary particles, electrons and scattered photons, produced when the primary photons interact. The primary interactions are first determined by computing the energy fluence incident on the patient representation and attenuating the primary intensity as the beam traverses through the patient representation.

The dose distribution is obtained by convolving the convolution kernels with the distribution of the total energy removed per unit mass (TERMA). The TERMA in the patient representation is the product of the mass attenuation coefficient and the energy fluence distribution. The convolution kernel represents the relative energy deposited per unit volume as a function of position where the primary photon interactions occur. The TERMA distribution accounts for the beam attenuation due to the surface contour and the composition of the irradiated material as well as inverse-square fall-off in beam intensity. The mass attenuation coefficient depends on the beam hardening to the point of interest and the material at the interaction point. The energy fluence is computed by ray tracing through the phantom along diverging paths with respect to the target. This is adequately accurate for photons originating from the target but does not exactly reflect the source positions of the extrafocal radiation. The failing is small if the dose contribution from extrafocal radiation is reduced and the scattered radiation outside the field is well approximated.

Basically, the convolution method is a blurring operation of the primary TERMA in the patient. The blurring is physically due to the Compton scatter and the electron transport from the site of primary photon interactions. A broad beam of an arbitrary shape is divided into narrow beams or pencils incident on the patient’s surface. The dose at a point of interest is computed by summing the contributions from each pencil. The pencil beam method has been employed for both photon and electron dose calculations. The dose distributions in a series of parallel planes normal to the central ray are calculated by the convolution of the two-dimensional profile of a pencil beam in each plane with the primary photon fluence. The convolution integral can be written as

\[
D(x, y, d) = \int \int \Phi(a, b, d)K(x - a, y - b, d)da db
\]

where

- \( D \) is the computed dose at depth \( d \) and at lateral coordinates \( x \) and \( y \) relative to the central axis
- \( \Phi \) represents the relative fluence distribution (TERMA) and includes the effects of all secondary field shaping devices, as well as the phantom
- \( K \) is the convolution kernel represented by the two-dimensional cross-sectional profile of the pencil beam at depth \( d \)
The pencil beam dose distribution used to compute kernel $K$ can be obtained by Monte Carlo calculations using the energy spectrum of the specific treatment machine or can be extracted from measurements of broad-beam profiles using a deconvolution method.

**Abbreviation:** TERMA = Total energy removed per unit mass.

### Coolidge tube

*(Diagnostic Radiology)* A Coolidge tube is an x-ray tube with a heated cathode, which replaced the gas discharge tubes used for x-ray production for the first two decades after Roentgen's discovery and investigation of the properties of x-radiation. These early tubes had several significant limitations.

The American physicist, William D. Coolidge, working in the research laboratories of the General Electric Company, led the development of a tube with improved characteristics. The tube was first introduced in 1913 and is the type used for x-ray production today.

A vacuum tube with a heated cathode, the Coolidge tube, provides several major advantages, including

- Much greater x-ray output
- Accuracy of adjustment
- Stability
- Flexibility in exposure factors
- Ability to duplicate exposure techniques
- Longer life of x-ray tube
- Absence of indirect radiation

### Cooling curve

*(Diagnostic Radiology)* See Anode cooling curve

### Coordinate system

*(General)* The coordinates in a coordinate system describe the location in a plane or space. For example, in two-dimensional medical imaging, two coordinates (often called x-coordinate and y-coordinate) give the position of an image pixel. The pixel position corresponds to a position on a detector and the pixel value is proportional to the amount of energy deposited in the specific detector element.

A coordinate system with three physical dimensions of space is called a 'Cartesian coordinate system'.

### Coordinate transformation

*(Nuclear Medicine)* Coordinate transformation refers to the process of transforming coordinates from one frame of reference to another. An example is the transformation between Cartesian coordinates and polar coordinates.

### Copper

*(General)*

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Cu</th>
</tr>
</thead>
<tbody>
<tr>
<td>Element category</td>
<td>Metal</td>
</tr>
<tr>
<td>Mass number $A$</td>
<td>63</td>
</tr>
<tr>
<td>Atomic number $Z$</td>
<td>29</td>
</tr>
<tr>
<td>Atomic weight</td>
<td>63.546 g/mol</td>
</tr>
<tr>
<td>Electronic configuration</td>
<td>$1s^2\ 2s^2\ 2p^6\ 3s^2\ 3p^6\ 3d^{10}\ 4s^1$</td>
</tr>
<tr>
<td>Melting point</td>
<td>1357.77 K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>2835 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>8.96 g/cm$^3$</td>
</tr>
</tbody>
</table>

### History:

For thousands of years, copper has been used by man in the making of jewellery, tools and decorative items. The earliest known copper artefact is estimated to originate from nearly 9000 BC. Early copper metal was obtained from naturally occurring samples, but the process of smelting ores of copper to obtain the pure metal was in global use by around 2000 BC.

### Isotopes of Copper:

Copper occurs naturally as one of two stable isotopes, $^{64}$Cu and $^{66}$Cu, which exist with a relative abundance of 69.15% and 30.85%, respectively. There are a further 17 synthetic radioactive isotopes, with atomic masses between 52 and 80. $^{64}$Cu and $^{66}$Cu are used in medical diagnostic imaging studies and may have potential uses in cancer therapies.

### Medical Applications:

**Germicide** – Copper is one of a number of metals that are toxic to living cells such as bacteria when in direct, prolonged contact. Copper metal is used in hospitals to limit the spread of germs, particularly in door handles and air conditioning systems.

**PET Imaging Tracer** – Two radioactive isotopes of copper are used as tracers in PET studies:

- $^{64}$Cu decays by emission of positrons (18%), beta particles (39%), electron capture (43%) and a small fraction of internal conversion and gamma emission. $^{64}$Cu can be produced by particle accelerator or nuclear reactor, and decays with a half-life of 12.7 h. When attached to an appropriate pharmaceutical, $^{64}$Cu can be used as a tracer in PET molecular imaging, and is currently being used to investigate perfusion, angiogenesis and hypoxia.
- $^{64}$Cu-ATSM is used in PET imaging, most commonly for myocardial perfusion imaging. $^{64}$Cu decays by positron emission (97%) with some electron capture, with a half-life of 9.7 min. A notable advantage of $^{64}$Cu is that it can be produced from its parent radionuclide, $^{64}$Zn, by generator.

### 64Cu Therapy

- In animal tests, $^{64}$Cu-labelled diacetyl-bis(N4-methylthiosemicarbazone (ATSM) has shown promise as a therapeutic agent for colorectal cancer. $^{64}$Cu-ATSM has been seen to target hypoxic cells and increases survival time of the animal without global toxicity. The short path length of beta emission allows significant damage to tumour cells with little collateral damage to healthy tissue. The ability to image the distribution of the agent using PET is likely to be advantageous.

### Related Articles:

- PET (positron emission tomography), Radionuclides in therapy

### Coronal plane

*(General)* To describe anatomical planes, imagine a person standing in an upright position and dividing this person with imaginary vertical and horizontal planes. Anatomical planes can be used to describe a body part or an entire body.

Think of a vertical plane that runs through the centre of your body from side to side at right angles to the sagittal plane. This plane divides the body into front (anterior) and back (posterior) regions. This plane runs through the central part of the coronal suture or through a line parallel to it; such a plane is known as a **coronal plane**.

### Related Article:

- Anatomical body planes

### Correction factor

*(Radiotherapy)* Correction factors are factors multiplied with a measured value in order to increase the accuracy of that measurement. An example is the correction factors used in quality control testing of linear accelerators used for external radiotherapy.

### Related Article:

- Precision
Correlation

(General) This term can either mean the statistical correlation coefficient between two variables, or the cross-correlation of two functions. For information about the latter, please see articles on Cross-correlation and Autocorrelation.

The correlation between two random variables indicates the strength and direction of a linear relationship. There are different types of coefficients that can be used depending on the situation.

Pearson’s Correlation Coefficient, \( r \): This is the most common of correlation coefficients, and ranges from +1 to −1. A correlation of +1 indicates a perfect positive linear relationship between variables, −1 indicates a perfect negative linear correlation, 0 indicates no linear relationship.

It is calculated by dividing the covariance between the two variables by the product of the standard deviations (std). Pearson’s correlation coefficient is calculated as follows:

\[
r = \frac{1}{n-1} \sum_{i=1}^{n} \left( \frac{X_i - \bar{X}}{\sigma_X} \right) \left( \frac{Y_i - \bar{Y}}{\sigma_Y} \right)
\]

where \( X_i, Y_i \) are measured values. Another commonly used metric is \( r^2 \), the coefficient of determination, which is equivalent to the proportion of variance explained by linear regression; i.e. if \( r^2 = 0.85 \), then 85% of the variability of \( Y \) can be explained by changes in \( X \) and a linear relationship.

It is important to note that the value of Pearson’s correlation coefficient only indicates the strength of a linear relationship between the variables; it does not give any indication of any other relationship (i.e. logarithmic, polynomial). Correlations for other relationships can sometimes be calculated by use of transformations to linear models (e.g. \( \ln(x) \) vs. \( y \)). Figure C.88 shows examples of non-linear relationships that are not identified with the use of Pearson’s correlation coefficient.

Pearson’s correlation coefficient also depends on the normality of the sample distributions. If this assumption is not valid, non-parametric methods should be used such as Spearman’s Rank Correlation Coefficient or Chi-Square. Normality can be tested with the use of a test such as the Kolmogorov-Smirnov test.

Spearman’s Rank Correlation Coefficient: This correlation coefficient, often denoted by \( \rho \), is a non-parametric measure of correlation, determining how well the relationship between two random variables can be described by a monotonic function. It is equivalent to the aforementioned Pearson’s correlation coefficient calculated from the ranks of the data, rather than the raw data itself.

Note that the correlation reflects the noisiness and direction of a linear relationship (top row), but not the slope of that relationship (middle), nor many aspects of non-linear relationships (bottom). N.B.: the Figure C.88 in the centre has a slope of 0 but in that case the correlation coefficient is undefined because the variance of \( Y \) is zero.

Related Article: Normal distribution

Correlation time

(Magnetic Resonance) The term correlation time \( \tau \), is used to describe the time over which a particular pattern in an experiment is correlated under the influence of a stochastic disturbance. Thus, it is the characteristic time over which fluctuations of a parameter occur. In MRI, the ‘correlation time’ concept is primarily used in connection with relaxation. In relaxation theory, the correlation time is the characteristic time between fluctuations in the local magnetic field (the lattice field) experienced by a spin due to the movements of neighbouring nuclei or molecules. The correlation time is often described as being the average time between molecular collisions, and is thus related to the binding state of the water molecules in the system. For example, free water molecules display rapid rotational movements, frequent collisions and short \( \tau \).

Related Articles: Relaxation, Relaxation rate, Relaxation time, B0, Temperature-dependence, Rotational averaging


Cost/benefit analysis

(Radiation Protection) Cost/benefit analysis must be used by an employer to justify any practice that involves the exposure of staff, patients or the public to ionising radiation. The benefits of the practice must outweigh the costs in terms of detriment or adverse radiation effects to those exposed.

For more information, see Justification.

Related Articles: Adverse radiation effects, Justification


Couch/patient

(Nuclear Medicine) The patient couch refers to the imaging table in an image modality system.

In transmission and emission imaging, the couch is typically made from a sturdy but low attenuating material so as not to produce any noticeable effect on the resulting images.

In MRI, the couch should be made out of non-magnetic material to prevent any image distortion.

Abbreviation: MRI = Magnetic resonance imaging.

Coulomb

(General) The coulomb (named after Charles-Augustin de Coulomb, a French Physicist) is defined as the amount of charge that flows through a given cross section of conductive wire in 1 s if there is a steady current of 1 A in the wire:

\[ q = It \]

where

- \( q \) is in coulombs
- \( I \) is in amperes
- \( t \) is in seconds


Related Articles: Charge, Force electrostatic

![Figure C.88](image-url)  Several sets of \((x, y)\) points, with the correlation coefficient of \( x \) and \( y \) for each set.
Coulombs/kg

(Radiation Protection) Coulombs per kilogram is the SI unit of exposure, given the name 'Roentgen'. More precisely, it is quantity of electric charge of one sign (+ve or –ve, measured in Coulombs) that is liberated by the interaction of ionising radiation and collected by the electrodes in an ionisation chamber containing a mass of air (in kilograms). 1 R = 1 C/kg.

Related Articles: Exposure, Roentgen (R)

Count rate

(Nuclear Medicine) The counting or count rate is the registered number of photons per unit of time. A high count rate is beneficial in many applications, for example imaging applications where high count rates enable fast examinations with adequate statistics. In some applications, high count rates can have negative effects, for example in a scintillation camera (see Apparent source position).

Related Articles: Apparent source position, Scintillation camera

Counting limitations

(Nuclear Medicine) Counting limitation describes a system’s capability to register and process very high event rates in the detector or detector system. Very frequent events will saturate the detection system and count losses arise due to dead time of the detector circuitry. The latter can be due to the type of the counting system: paralyzable; non-paralyzable or a combination of both.

A scintillation camera can have very severe counting limitations if the pile-up rejection is not working and mispositioning of count events can occur at very high count rates. It also has to be remembered that it is the total number of photons impinging on the crystal across the whole energy spectrum that is of importance.

Related Articles: Count losses, Pile up, Paralyzable counting system, Non-paralyzable counting system


Counting systems

(Nuclear Medicine) A counting system is a detector that measures the activity in radioactive samples. The most common counting system is the well counter.

Related Article: Well counter

Counting times

(Nuclear Medicine) See Acquisition time

Coupling efficiency

(Ultrasound) Coupling efficiency describes the efficiency with which the ultrasound wave from the active transducer element is transmitted into the tissue medium. The design of the transducer incorporates matching layers of optimal thickness and acoustic impedance to transmit as much energy across the bandwidth as possible. High coupling efficiency is also important for the returning echoes.

Related Articles: Backing material, Acoustic impedance, Transducer

Coupling medium

(Ultrasound) Gel is most commonly used as coupling medium in diagnostic examinations, Figure C.89. The purpose is to avoid air to be trapped between the transducer and the skin. As air has much lower acoustic impedance than soft tissue, 400 Rayls compared to 1.6 × 10⁶ Rayls, total reflection will occur at the boundary with trapped air, which will reduce the image quality markedly.

Oil or water can also be used as coupling medium.

Related Articles: Coupling efficiency, Reflection coefficient, Rayl

CP (circularly polarised)

(Magnetic Resonance) See Circularly polarised (CP)

CPMG (Carr–Purcell–Meibom–Gill)

(Magnetic Resonance) See Carr–Purcell–Meibom–Gill (CPMG) sequence

CR (computed radiography)

(Diagnostic Radiology) See Computed radiography (CR)

Creatine

(Magnetic Resonance) Creatine (Cr) is a chemical compound that features in in vivo proton (1H) NMR spectra of a number of organs (Figure C.90).

Protons in the methyl group in creatine and phosphocreatine give rise to a resonance at 3.02 ppm (thus, these compounds cannot be distinguished in 1H MRS). There is a further resonance at 3.9 ppm from CH₃ protons, but as this is smaller and obscured by overlapping peaks, it is normally considered less important.

Creatine plays a role in the body’s energy metabolism. Buffering mechanisms ensure that its concentration in the normal brain is usually stable, and therefore it is often used as an internal reference for

CH₃ O
NH = C – N – CH₂ – C – OH
NH₂

FIGURE C.90 Molecular structure of creatine.
quantitative spectroscopy. However, it is reduced in certain types of tumours and slightly increased in reactive gliosis.

The importance of phosphocreatine in 31P NMR arises from its role in the body’s energy metabolism. PCr provides a readily available source of phosphorus to support generation of ATP from ADP, and is depleted to maintain ATP levels during ischaemia and hypoxia. Thus, PCr levels can be used to monitor the effect of fatiguing exercise in skeletal muscle. It also provides a prognostic indicator in birth asphyxia that is well correlated with neurodevelopmental outcome (Figure C.91).

**Related Article:** Phosphocreatine

### Critical structures

**Radiotherapy** The goal of radiotherapy is to deliver as high a dose as possible to the tumour while sparing surrounding normal tissue. Certain organs within the body are very sensitive to radiation; hence, the dose delivered to the tumour is limited. These organs are called critical structures or organs at risk. They must be taken into consideration during the treatment planning process. Some examples of critical structures are the lungs, the spinal cord, the eyes, the rectum and the bladder. Absorbed dose to the critical structures and the percentage of irradiated volume is limited. These limits are called constraints. DVHs or NTCPs can be used for plan evaluation from the point of view of dose to the critical structures. Therefore, these critical structures near to the tumour limit the amount of radiation that can be administered due to their dose tolerances.

**Abbreviations:** DVH = Dose volume histogram and TCP = Normal tissue complication probability.

**Related Articles:** Organ at risk, Conformal radiotherapy, Custom blocking, Normal tissue complication probability, Normal tissue dose, Normal tissue dose response, Normal tissue reaction, Normal tissue toxicity, Tolerance

### Crookes tube

**Diagnostic Radiology** The Crookes tube is a partially evacuated vacuum discharge tube with electrodes in a glass envelope. The tube had been developed by Sir William Crookes, an English chemist and physicist, to study several electrical phenomena. Such type of tube has been used by W.C. Roentgen when he discovered the x-rays.

**Further Reading:**

### Cross-correlation

**Radiation Protection** The concept of cross section, symbol, \( \Phi \), is used to describe the probability or likelihood of the interaction of charged or uncharged particles, or photons, with nuclei or other ‘target entity’. If a beam is interacting with a target, then the cross section is the average area perpendicular to the direction of the radiation, which has to be assigned to each nucleus in order to account geometrically for the total number of interactions that occur across an incident beam of given fluence. The unit of cross section is therefore that of area (m²). However, although it can be thought of as the ‘area of influence or interaction’, or as the ‘target area’ afforded by each nucleus/atom, it must be considered as completely independent of the actual physical dimensions of the nucleus/atom.

The cross section depends not only on the type of the target but on the type and energy of the particles. Cross section is not limited to beams – it can expressed in more general terms and the term is defined by the International Commission on Radiation Units and Measurements (1998) as follows:

\[
\sigma = \frac{P}{\Phi}
\]

Unit: m²

Fluence. The fluence, \( \Phi \), is the quotient of \( dN \) by \( da \), where \( dN \) is the number of particles incident on a sphere on a sphere of cross-sectional area \( da \), thus

\[
\Phi = \frac{dN}{da}
\]

Unit: m⁻²

The barn, b, is a special SI unit of cross section:

\[
1 \text{ barn} = 10^{-28} \text{m}^2 = 100 \text{ fm}^2
\]

**Related Article:** Barn

**Further Reading:** Barn

### Cross-correlation

**Ultrasound** In signal processing, the cross-correlation is a measure of how similar two signals are. For instance, it can be used to find features in an unknown signal by comparing it to a signal with desired shape. The cross-correlation is similar to convolution of two signals, but does not involve a time-reversal, only a shift and multiply. The autocorrelation is a cross-correlation, but with the signal itself. Consequently, the autocorrelation will always have a peak at zero lag.

Cross-correlation in diagnostic ultrasound is used as an alternative to the autocorrelation algorithm to estimate blood flow velocity. The resulting estimate is a time-shift, which is a measure of how much a scattering target has moved between two transmit pulses. Cross-correlation has a number of advantages, foremost the ability to detect higher velocities without aliasing. The same pulse can also be used to form the B-mode image as for flow velocity estimation. The drawback is however an uncertainty in the estimation.

Each RF line is divided into segments, where each segment is correlated to the same segment of the subsequent RF line. The

---

**Figure C.91** ¹H NMR spectrum of the human brain showing Cr resonance at 3.02 ppm.
estimate can be improved by averaging correlations from several RF lines. Usually, the smallest detectable velocity (corresponding to one sample point) is too coarse; so one method to improve the resolution is to fit a second-order polynomial to the three points at the peak, and then find the peak position from the polynomial.

If a full calculation of the cross-correlation is performed, this will amount to a very large number of calculations. To reduce the computational load, the calculation may only need to be performed by converting the RF data to one-bit signals, i.e. 1 for the positive part of the signal, and −1 for the negative. For an infinite number of samples, this method is exact in finding the time shift, but as the data are limited, there will be a deviation from the ‘true’ time shift, however normally very small.

**Related Article:** Autocorrelation

### Cross-hairs

**(Radiotherapy)** Cross-hairs are lines and marks projected onto the image obtained on a simulator in order to visualise the treatment field with respect to the patient’s anatomy and to ensure the correct treatment geometry. The treatment field can be centred using the central cross of the cross-hairs (see Figure C.92). In addition, cross-hairs can be used to show the extent of the treatment field and usually have a scale or graticule graduated in centimetre intervals in order to allow measurements to be made with respect to the centre or field edges, etc.

**Related Article:** Simulation

### Cross-line curves

**(Radiotherapy, Brachytherapy)**

**The Paris Dosimetry System:** For the Paris System dose calculations, dose rates in water at the central plane of a linear source are needed, where oblique filtration and attenuation and scattering in water are taken into account.

Besides tabulated values and Escargot curves, so-called cross-line curves/graphs are also used. Cross-line curves give the dose rate for 192Ir-wires of unit linear source strength as a function of distance from the centre of the wire, in the central plane and also in planes parallel to the central plane. One graph is used for each 192Ir wire length.

**Related Articles:** Paris system, Escargot curves

**Further Reading:** Venselaar, J. and J. Pérez-Calatayud, eds. 2004. A Practical Guide to Quality Control of Brachytherapy Equipment, ESTRO Booklet No 8, European Society for Therapeutic Radiology and Oncology, Brussels, Belgium.

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**Crossed grid**  
(Diagnostic Radiology) See Grids, crossed

**Crosstalk**  
(Magnetic Resonance) Crosstalk is an artefact caused by the imperfect shape of RF pulse slice profiles. An ideal RF pulse slice profile should be square, but in reality they are typically more bell-shaped. In multi-slice mode with contiguous slices, an RF pulse can therefore partially excite adjacent slices. Crosstalk can reduce the signal intensity and/or modify the image contrast through partial saturation.

The artefact occurs when slices are too close together and can be eliminated by ensuring the spacing between slices is typically at least 10% of the slice thickness. 3D imaging can be used if it is important to view the whole volume.

**Abbreviation:** RF = Radio frequency.

**Related Articles:** Slice profile, Sli ce spacing


### CRT (cathode ray tube)

(Diagnostic Radiology) See Cathode ray tube

**Cryogen**  
(Magnetic Resonance) Cryogen is the cooling agent that is used to maintain a superconducting magnet at a sufficiently low temperature. Normally, liquid helium (4.2 K) is used. The cryogen will tend to boil off and needs to be refilled periodically. Many systems now incorporate cryo-compressors to liquify gaseous helium. Since helium is expensive, extensive boil off is prevented by the cryostat (see related article), minimisation of heat loss pathways and for older MR systems by the use of liquid nitrogen (77 K) as a second cryogen, providing a thermal shield to the more expensive helium. Increasingly, cryorefrigerators are being used in smaller superconducting magnets, reducing the need for cryogen replenishment (Figure C.93).

The cryogen (helium) surrounds the superconductive magnet coil.

**Related Article:** Cryostat

### Cryostat

**(Radiation Protection)** A cryostat is a device for maintaining a constant very low temperature, i.e. below 123 K. The use of a cryostat is indispensable for many semiconductor radiation detectors (germanium or silicon) that are required to operate at the temperature of liquid nitrogen (77 K), or in magnetic resonance imaging for maintaining the static field superconducting magnet (∼2 K).

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![Figure C.92](image)

**Figure C.92** Cross-hairs with centimetre scale or graticule and superimposed field of 11 × 8 cm.

![Figure C.93](image)

**Figure C.93** Schematic cross-sectional drawing of a MRI system.
There are many kinds of cryostats; the simplest cryostat is a Dewar vessel in which the thermal insulation minimises heat flows by conduction, convection or radiation from and to the environment. There are two types of these vessels made from glass or stainless steel double walls, containing a vacuum or a combination of vacuum and super-insulation inside. The Dewar containing liquid helium (boiling temperature 4.2 K) is surrounded by a second one containing liquid nitrogen (77 K). Dewars can be made in many shapes and sizes from simple cylindrical vessels to large tanks.

**Related Article:** Semiconductor detector


**C-scan**

(Ultrasound) The term C-scan describes a scan to produce an image in a plane parallel to the transducer face. For a transducer placed on the skin, the resulting image is parallel to the skin surface. Compared to a conventional B-mode image, C-scans must obtain echoes from an axis orthogonal to the B-mode plane. This can be done by moving a transducer in the elevation plane or using a 3D volume acquisition (e.g. from a 2D array), as illustrated in Figure C.94.

**CSI (chemical shift imaging)**

(Magnetic Resonance) See *Chemical shift imaging (CSI)*

**CT (computed tomography)**

(Diagnostic Radiology) See *Computed tomography (CT)*

**CT dose index**

(Diagnostic Radiology) See *CTDI*

---

**CT Fluoroscopy**

(Diagnostic Radiology) CT fluoroscopy (CTF) is a mode of imaging, which offers a continuously updated image. In CTF, the couch is not incremented through the gantry in the manner of sequential or helical scanning, but the same volume of the patient is viewed over a period of time in a manner analogous to fluoroscopy in conventional radiology. The main use of CTF is in interventional procedures such as tissue biopsies and fluid draining (Figure C.95).

The main requirements for CTF are a continuously rotating gantry (as enabled by slip-ring technology), and fast collection and reconstruction of data; so there is only a short delay between data collection and image display. The first image in the series is reconstructed from the first rotation. The second image is reconstructed by subtracting the first segment of data, for example the first 60°, and adding data from first 60° of the second rotation. Subsequent images are reconstructed by always subtracting the oldest 60° of data and replacing it with the most recently acquired 60°. In this way, six images per gantry rotation (360/60) can be displayed for each simultaneously acquired z-axis position.

In order to provide faster display of images, reconstruction is usually performed initially on a 256 × 256 matrix, rather than the conventional 512 × 512.

When used for interventional procedures, additional hardware is usually required so that the CT scanner can be operated from inside the scanner room. This includes a scan room monitor and a console with a foot switch for controlling exposure and a joystick, or similar, for couch-top movement.

Radiation dose to the patient should be closely monitored in CTF. Although, generally, lower tube currents are employed than in conventional CT scanning, the same area of the patient is irradiated continuously and this can result in high local organ doses. When operating in CTF mode, an alarm is employed to alert the operator after a preset time limit, for example if 100 s have been exceeded.

When performing procedures from within the scan room, appropriate precautions should be taken by the operator and doses to the hands, eyes and body carefully monitored to ensure limits are not exceeded. Operators should wear protective aprons, when practical use needle holders to keep the hands out of the direct beam, and stand as far away as possible from the scan plane during scanning.

**CT number**

*(Diagnostic Radiology)* The CT number is the numerical value of each pixel in a CT image matrix, and is proportional to the linear attenuation coefficient of the material in a voxel represented by the particular pixel. Modern diagnostic scanners use Hounsfield units (HU) to express CT numbers. On the Hounsfield scale, attenuation coefficients are normalised to that of water, and the CT number of a material, \( m \), is given by the following formula:

\[
\text{CT number } (m) = \frac{\mu_m - \mu_{\text{water}}}{\mu_{\text{water}}} \times 1000 \text{ HU} \quad (C.10)
\]

where \( \mu_m \) and \( \mu_{\text{water}} \) are the linear attenuation coefficients of a material, \( m \) and water, respectively. From Equation C.10, the CT number of water is 0 HU, and, as the attenuation of air can be approximated to zero, that of air is −1000 HU.

CT images are generally displayed on a greyscale, where more attenuating materials are represented by lighter shades and the less attenuating by darker ones. The standard Hounsfield scale ranges from approximately −1000 to +3000 HU, so contains \( 2^{12} \) levels of grey (Figure C.96a). The human eye however is only capable of distinguishing between about 1000 levels of grey on the monitor.

The range of CT numbers displayed on the monitor can be varied by adjusting the window level (WL) and window width (WW), according to the tissues of interest. Better differentiation between tissues can be perceived if the WW is reduced, so that a smaller range of HU is displayed over the entire greyscale (Figure C.96b and d). The WL can then be set to display the tissues of interest (Figure C.97).

Typical CT numbers of tissues, on the Hounsfield scale, are

<table>
<thead>
<tr>
<th>Tissue</th>
<th>Approx. CT number (HU)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lung</td>
<td>−550 to −950</td>
</tr>
<tr>
<td>White matter</td>
<td>30</td>
</tr>
<tr>
<td>Grey matter</td>
<td>40</td>
</tr>
<tr>
<td>Muscle</td>
<td>50</td>
</tr>
<tr>
<td>Fat</td>
<td>−90</td>
</tr>
<tr>
<td>Trabecular bone</td>
<td>300–500</td>
</tr>
<tr>
<td>Cortical bone</td>
<td>600–3000</td>
</tr>
</tbody>
</table>

**Related Articles:** Window, Hounsfield number, Hounsfield scale

![FIGURE C.96](Figure Caption)

**FIGURE C.96** The CT number scale in Hounsfield units: (a) the full range displayed over the greyscale; (b) WL changed and WW reduced; (c) WW reduced further to display a smaller range of CT numbers over the greyscale; and (d) WL changed to display different range of CT number values.

![FIGURE C.97](Figure Caption)

**FIGURE C.97** Window level and WW set to display (a) lung tissue and (b) pulmonary metastases.
**CT reconstruction**  
*(Diagnostic Radiology)* The history of tomographic image reconstruction goes back to the early twentieth century. Austrian mathematician Johann Radon (1887–1956) developed the mathematical basis of reconstructing tomographic images from sinograms (see images given in the following, in the article on Back-projection and also references) containing multiple attenuation profiles collected around the object. Wider use of this method in medical use became applicable with the development of computers and computed tomography (CT).

Profiles are generated by detecting the thinly collimated x-rays transmitted through the object (patient). The sinogram consists of views (number of shots done during one rotation) and rays (number of elements in one shot). The number of views (~1000 at modern scanners) affects the circumferential component of resolution, and the number of rays (~800/view), the radial component of resolution in the CT image. The task is to compute from the preprocessed sinogram data the linear attenuation coefficient *(Equation C.11)* for each pixel in the image (2D) or each voxel of the object (3D):

\[
\mu = \ln \left( \frac{I_0}{I} \right)
\]  
*(C.11)*

These values are normalised and scaled to present CT numbers in Hounsfield units (HU), *(Equation C.12)*. The CT numbers are further to be presented as a greyscale image:

\[
\text{CT(HU)} = 1000 \times \frac{(\mu_{\text{tissue}} - \mu_{\text{water}})}{(\mu_{\text{water}} - \mu_{\text{air}})}
\]  
*(C.12)*

The methods used for CT reconstruction include back-projection method, iterative reconstruction and analytical methods. Only the latter category method, i.e. filtered (or convolution) back-projection method is nowadays applied. A CT reconstruction demo with a few choices of different parameters is shown in Figure C.98.

**Related Articles:** CT, Back-projection, Iterative reconstruction, Filtered back-projection


**CT scanner**  
*(Diagnostic Radiology)* See Computed tomography

**CT simulator**  
*(Radiotherapy)* The major steps in the target localisation and treatment field design are as follows:

- Acquisition of the patient data set
- Localisation of target and adjacent structures
- Definition and marking of the patient coordinate system
- Design of treatment fields
- Transfer of data to the treatment planning system (TPS)
- Production of an image for treatment verification

CT simulators are CT scanners equipped with special features that make them useful for certain stages in the radiotherapeutic process. The special features typically are as follows:

- A flat table top surface to provide a patient position during simulation that will be identical to the position during treatment on a megavoltage machine.
- A laser marking system to transfer the coordinates of the tumour isocentre, derived from the contouring of the CT data set, to the surface of the patient. Two types of laser marking systems are used: a gantry-mounted laser and a system consisting of a wall-mounted moveable sagittal laser and two stationary lateral lasers.
- A virtual simulator consisting of software packages that allow the user to define and calculate a treatment isocentre and then simulate a treatment using digitally reconstructed radiographs (DRRs).

In CT simulation, the patient data set is collected and target localisation is carried out using CT images with fluoroscopy and radiography replaced by DRRs. A laser alignment system is used for marking and a virtual simulator software package is used for field design and production of verification images. Transfer of all necessary information to the TPS is achieved electronically.

A CT simulator essentially obviates the need for conventional simulation by carrying out two distinct functions:

- Physical simulation, which covers the first three of the six target localisation steps listed earlier
- Virtual simulation, which covers the last three of the six target localisation steps listed earlier

**Related Article:** DRR


**CT x-ray tube**  
*(Diagnostic Radiology)* Basic operation of the CT x-ray tube is similar to that of conventional radiology. Typically, two (‘small’ and ‘large’) focal spots are available, which are automatically selected according to the protocol. Total filtration is typically 6–10 mm Al equivalent. The heat capacity and the cooling rate of the tube have been designed for long acquisition times and heavy current loads, typical values are 5–8 MU and 1500–5000 kHU/min, accordingly. The tube assembly in the gantry has been designed and has to be balanced carefully for each installation to allow the fast rotation times (<1 s) used in diagnostic imaging.

Although the new CT x-ray tubes allow for dramatic increase of the power and intensity of the exposure, the focal spot is still determined by the size of the cathode filament. A new x-ray tube (commercial name STRATON) has been developed by SIEMENS, which solves this problem *(Figure C.99)*. This tube has a completely different construction. It uses thermal electrons created by the filament, but further applies electromagnetic deflection and focusing of the beam of thermal electrons, this way making the focal spot size independent of the filament size. In order to dissipate the high thermal energy imparted to the anode, this tube construction uses a stationary round anode, sealed with the tube metal envelope *(Figure C.100)*. This way the whole assembly (envelope plus anode) rotates. This allows for leaving the backside of the anode (and the bearings) to be directly open and cooled by the insulating oil of the tube.
The power of STRATON is very high; the focal spot very small; the cooling time very short (down to 20 s with cooling rates of 4.7 MHU/min).

A new x-ray tube developed by Philips (iMRC) also uses external focusing of the beam of thermal electrons from the cathode, but the large rotating anode is inside the metal envelope. The power of this new tube reaches 120 kW.

Most new CT x-ray tubes allow automatic tube current modulation (ATCM) – adjusting the x-ray tube current according to the variations in patient attenuation (see the eponymous article).

**Related Articles:** X-ray tube, CT scanner, Automatic tube current modulation

**Further Readings:**
Absorbed dose in CT is usually expressed in terms of the CTDI, computed tomography dose index. The unit of CTDI is gray, but doses are generally at the level of units to tens of milliGray (mGy). The CTDI represents the absorbed dose from a series of scanner rotations, although the dose is traditionally measured with a single axial scan, from which the series dose is calculated using the following formula:

$$\text{CTDI} = \frac{1}{nT} \int_{-\infty}^{+\infty} D(z)dz$$

where

- $D(z)$ is the dose profile along the $z$-axis (scanner axis of rotation)
- $T$ is the nominal width of single acquired slice
- $n$ is the number of slices acquired per rotation

Therefore

$$nT$$

is the nominal collimated $z$-axis x-ray beam width

The preceding equation represents the general form of CTDI. A number of specific CTDI definitions have been formulated over the years, and currently the most commonly used is the $\text{CTDI}_{100}$. $\text{CTDI}_{100}$ expresses the absorbed dose for a scanned length of 100 mm and is calculated as follows:

$$\text{CTDI}_{100} = \frac{1}{nT} \int_{-50}^{+50} D(z)dz$$

$\text{CTDI}_{100}$ is a strictly defined term. It is measured in standard, cylindrical, polymethyl methacrylate (PMMA) phantoms (Figure C.101). Both phantoms are a minimum of 14 cm long and have diameters of 16 and 32 cm, to represent a head and a body, respectively. Originally, CTDI measurements were usually made with thermoluminescent (TLD) dosimeters, but currently, specialist 100 mm long pencil ionisation chambers are most commonly used (Figure C.102). The absorbed dose is quoted as dose to air; therefore, for a dosimeter calibrated in air kerma, the dosimeter reading can be used directly, and the integral of the dose profile over 100 mm obtained by taking the product of the dosimeter reading and the active chamber length. The standard CT dose phantoms have holes at appropriate locations, to allow placement of the dosimeter (pencil ionisation chamber). Measurements are made at the centre of the phantoms, $\text{CTDI}_{100,c}$ and at the periphery, at a depth of 10 mm, $\text{CTDI}_{100,p}$.

The average absorbed dose in the scan plane of the phantom is expressed by taking a weighted average of centre and periphery $\text{CTDI}_{100}$ measurements. This weighted average, $\text{CTDI}_{w}$, is calculated using the following formula:

$$\text{CTDI}_w = \frac{1}{3} \text{CTDI}_{100,c} + \frac{2}{3} \text{CTDI}_{100,p}$$

The preceding equation represents the absorbed dose for a contiguous scan, i.e. a sequential (axial) scan where the table increment between rotations is equal to the nominal beam width, or a helical (spiral) scan with a pitch value of 1. For non-contiguous scanning, the $\text{CTDI}_w$ can be adjusted to give the volume CTDI, $\text{CTDI}_{vol}$, using the following formula:
Cumulated activity

(Nuclear Medicine) When trying to determine radiation doses to patients in therapeutic or diagnostic nuclear medicine applications, one uses the cumulated activity $\bar{A}$. $\bar{A}$ is a measure of the total number of disintegrations in a source organ between initial uptake and total clearance. The cumulated activity depends on two kinetic properties of the used radio-compound, i.e. (1) how much and how fast is the activation accumulated in each source organ and (2) how long it takes until all activity is cleared. The cumulated activity is the product of these two factors and the SI unit is ‘Bq × s’. In other words, the cumulated activity is the total number of disintegrations in the source organ between initial accumulation and total radionuclide clearance.

The temporal and spatial distribution of a radiotracer in the body depends on a number of radio-compound properties: delivery, uptake (accumulation), metabolism, clearance, excretion and physical decay. After injecting activity, the concentration of radionuclides (i.e. activity) in a source organ will change over time, which is described by a time-activity curve $A(t)$. If the time-activity curve is known, then the cumulated activity equals the area under the curve, i.e. an integration over $A(t)$ from $t = 0$ to infinity:

$$\bar{A} = \int_{0}^{\infty} A(t)dt$$  \hspace{1cm} (C.13)

$t = 0$ represents the time for activity administration. As previously mentioned, the cumulated activity depends on a number of factors and the time-activity curve is unique for each source organ and radionuclide. Four scenarios are mentioned in the following text, where the cumulated activity is estimated.

**Scenario 1:** The uptake in the source organ is near-instantaneous and there is no biological excretion (i.e. only a physical decay component). Cumulated activity is then described by an exponential radioactive decay function. The cumulated activity for such a scenario is

$$\bar{A} = 1.44T_pA_0$$  \hspace{1cm} (C.14)

where
- $T_p$ is the physical decay constant
- $A_0$ is the activity initially present in the organ

**Scenario 2:** The uptake is near-instantaneous and the biological half-life $T_b$ is very short compared to the physical decay, i.e. biological excursion only. The biological decay function often consists of several exponential excursion components with a fraction $f_1$ of the initial activation $A_0$ being excreted with half-life $T_{b1}$ and fraction $f_2$ being excreted with half-life $T_{b2}$ and so on. In such a scenario, the cumulated activity is given by

$$\bar{A} = A_0\int_{0}^{\infty} f_1e^{-0.693T_{b1}t}dt + A_0\int_{0}^{\infty} f_2e^{-0.693T_{b2}t}dt + \cdots$$  \hspace{1cm} (C.15)

**Scenario 3:** Near-instantaneous uptake with a significant contribution from both biological excursion and physical decay. If assumed that the biological half-life $T_b$ is described by a single-component exponential decay, the physical and biological half-life can be expressed as an effective half-life $T_e$ given by Equation C.16a. For such a scenario, the cumulated activity is given by Equation C.16b:

$$T_e = \frac{T_pT_b}{(T_p + T_b)}$$  \hspace{1cm} (C.16a)

$$\bar{A} \approx 1.44T_eA_0$$  \hspace{1cm} (C.16b)

If the biological half-life is a multi-exponential decay, the effective half-life for each component is given by Equation C.16a. The cumulated activity is calculated with the effective half-lives replacing the biological half-life in Equation C.15.

**Scenario 4:** When the uptake is not near-instantaneous, i.e. when a significant amount of the activity decays before accumulating in the organ, the equations mentioned earlier will overestimate the organ doses. In most cases, the uptake can be described by

$$A(t) = A_0(1 - e^{-0.693T_bt})$$  \hspace{1cm} (C.17)

where $T_b$ is the biological uptake half-time. In such a scenario, the cumulated activity is given by

$$\bar{A} = 1.44A_0\left(\frac{T_{u\infty}}{T_b}\right)$$  \hspace{1cm} (C.18)

where $T_{u\infty}$ is the effective uptake half-time derived from

$$T_{u\infty} = \frac{T_pT_b}{T_u + T_p}$$  \hspace{1cm} (C.19)

**Related Articles:** MIRD formalism, Equilibrium absorbed dose constant, Absorbed fraction, Mean dose per cumulated activity

Cumulative dose

(Radiation Protection) The total dose (defined in terms of either the absorbed dose, equivalent dose or effective dose) that would be received by an individual from repeated exposures to ionising radiation over a period of time is called the cumulative dose. This is distinct from the committed dose received over time from a single injection, inhalation or ingestion of a radioactive substance.

Related Articles: Absorbed dose, Equivalent dose, Effective dose, Committed dose

Cumulative dose volume histogram

(Radiotherapy) See Dose volume histogram

Curie

(Radiation Protection) Curie (Ci) is an old unit of radioactivity, defined as

\[ 1 \text{ Ci} = 3.7 \times 10^{10} \text{ decays/s} \]

Originally, the curie was chosen as roughly the activity of 1 g of the radium isotope 226Ra.

The SI unit of activity (radioactive decay) is becquerel (Bq), which is to one decay per second, this way

\[ 1 \text{ Ci} = 3.7 \times 10^{10} \text{ Bq} \]

The curie is named after Pierre Curie and Marie Sklodowska Curie.

Related Article: Becquerel

Current consumption

(General) Current consumption is the current that an electric circuit draws from the voltage or current source.

Related Article: Current source

Current, eddy

(Magnetic Resonance) See Eddy currents

Current intensity

(General) This term normally refers to the intensity of the anode current (tube current) in an x-ray tube. Current intensity is directly related to the density of the thermal emission current (see articles on Filament heating and Tube current). The current intensity (the number of electrons per unit area) is measured in A/mm² but, as the source of these thermal electrons (the cathode filament wire) is with standard size for a particular x-ray tube, simply mA is used in practice. Current intensity is linearly related to the intensity of the x-ray photons in the beam.

Related Articles: Filament heating, Tube current

Current source

(General) A current source is an electrical or electronic device that delivers or absorbs electric current where the current is independent of the voltage across it, in the ideal case. The ideal current source has infinite internal resistance where real current sources can be represented as ideal current sources in parallel with a resistance. The simplest current source consists of a voltage source in series with a resistor.

Curvature correction

(Radiotherapy) Patient curvature can be important when planning treatments with electron fields. This can be illustrated by considering the set up shown in Figure C.103, where a single electron field is used to treat a chest wall area. Two factors affect the dose just under the skin surface – the increase in SSD at different parts of the beam as it enters the patient, and also the effects of oblique incidence.

With electron treatments, oblique incidence can give rise to electrons depositing dose non-uniformly across the field area (see Figure C.103). In region ‘a’, the initial direction of travel of the electrons is perpendicular to the skin surface whereas this is not the case in regions ‘b’ and ‘c’, and the obliquity in these regions will result in a relatively higher dose there than at points in region ‘a’. This is because the electrons are effectively side scattered to a higher degree in the tissue instead of forward scattered as in region ‘a’. This is an effect of patient curvature and becomes greater as the obliquity becomes greater. As the angle of obliquity increases due to curvature, it is often best to arrange the angle of incidence of the central axis so that the angles of obliquity at the two extreme field borders in this case ‘b’ and ‘c’ are about equal if possible.

The problem of oblique incidence caused by patient curvature is also important in electron arc therapy. Efforts are made to minimise this by placing the linear accelerator isocentre (the rotational axis) as close as possible to the centre of curvature of the region to be treated.

Related Articles: Electron arc aperture, Electron oblique incidence


Curvilinear array transducer

(Ultrasound) Curvilinear arrays, also referred to as convex arrays, are array transducers in which the elements, typically 64, 128 or 256 are arranged in an arc (Figure C.104). This format allows for a large field of view from a small contact point (Figure C.105). The
transducer is designed depending on its application; the main factors are length of the array, radius of curvature and frequency. The width of the elements also dictates its performance; the aperture in the elevation plane is designed to focus at particular depth (Figure C.106).

Tightly curved arrays are used in intracavity probes (link) to provide large fields of view from small contact areas (Figure C.107).  

**Related Article:** Convex array

### Custom blocking

_{Radiotherapy}_ Some treatment units have only collimators that form rectangular fields (not multileaf collimators). Since treatment volumes are rarely rectangular, high-density shielding blocks are used to protect normal tissue and critical structures within the irradiated area. The blocks are either individually designed blocks fabricated from a low-melting-point alloy (custom blocking) or standard (library) blocks that may be purchased from the vendor of the treatment machine. All these blocks are placed on a plastic tray to correctly position them within the radiation field.

Customised blocks are fabricated as follows. The area to be blocked is drawn on a radiograph or exported directly from TPS. A block cutting machine cuts the shape of the block in a piece of Styrofoam. This mould is filled with low-melting-point alloy. Then the block is removed from the Styrofoam. Customised blocks follow the beam divergence (see Figure C.108).

Block transmission factors and tray transmission factors must be known for treatment planning.

Other solutions to customised blocks are possible, such as the computerised milling blocks of lead directly from shapes drawn of radiographs.

The shape of the shielded area is drawn on a radiograph (a). The mould shape is cut by moving the end of the hot wire along the line drawn on the radiograph (b). Low-melting-point alloy in liquid form is poured into the mould (c) and allowed to cool and set. The Styrofoam is then removed leaving the block (d) and this can then be placed on the linear accelerator (linac) blocking tray (e).

**Abbreviation:** TPS = Treatment planning system.

**Related Articles:** Critical structures, Organ at risk, Conventional radiotherapy, Block transmission factor, Low-melting-point alloy


### Cut film changer

_{Diagnostic Radiology}_ Another name for a type of spot-film camera.

**Related Article:** Spot-film camera

### Cutoff frequency

_{Nuclear Medicine}_ In nuclear medicine, the cutoff frequency typically refers to a frequency in a filter where frequencies on one side of the cutoff frequency are weighted in order to enhance certain features in a filtered image. Generally, image filtering is performed in Fourier space. In Fourier space, the image data are represented as a series of sine and cosine functions, where the high-frequency components represent high contrast differences, for instance at boundaries between bone and tissue, while low-frequency components represent the overall image intensity pattern with low contrast differences. In a low-pass filter, for example frequency components over a certain frequency threshold are weighted to decrease the signal to noise ratio.

**Related Articles:** Ramp filter, Signal to noise ratio (SNR)

### CW Doppler

_{Ultrasound}_ CW Doppler is short for continuous wave Doppler. The Doppler shift arises when there is a relative motion between...
the transmitter and receiver of sound. In diagnostic ultrasound, there is no relative motion between these two, but when the emitted sound is reflected off a surface, or scattered by particles that are moving, there will be an apparent shift in wavelength, and thereby frequency, which is dependent on the relative velocity between the surface/particles and the transducers.

**CW Doppler** refers to the principle of emitting a continuous frequency (or tone) with a single transducer, and receiving on a second transducer. As these two cannot be located at the same position, there is bound to be an angle between them, and thereby an overlapping zone between their respective beam patterns. This overlapping zone is referred to as the sample volume from where Doppler shifts can arise.

It can be shown that the resulting Doppler shift frequency, the difference between the received and transmitted frequencies, obeys the following relation when the velocity of the target is much smaller than the speed of sound:

$$\Delta f = \frac{2vf_0}{c_0} \cos \theta$$  \hspace{1cm} (C.20)

where
- $\Delta f$ is the Doppler shift frequency
- $v$ is the velocity of the target
- $f_0$ is the transmitted frequency
- $c_0$ is the speed of sound
- $\theta$ is the angle between the transmitter/receiver combination and the velocity of the particle

This is known as the Doppler equation and also assumes that the angle between the transmit and receive beams is small.

A practical implementation of a CW Doppler system is shown in **Figure C.109**, where the received signal is mixed with the transmitted frequency. Thus, both sum and difference frequencies are obtained, where the difference frequency corresponds to the Doppler shift as given by the Doppler equation. In the case depicted here, the low-pass filter may actually be omitted since the frequency response of the audio amplifier is much below the sum frequency produced in the mixer. If the direction of the target (i.e. the sign of the Doppler shift) is required, a quadrature demodulation of the received signal is necessary.

**Related Article:** CW (continuous wave)

**Cyberknife**  
(Radiotherapy) This machine manufactured by Accuray delivers multiple highly focused beams of radiation using a robotic arm capable of delivering hundreds of uniquely angled beams in a single fraction (see Linear accelerator, Figure L.23). The system uses a 3D, non-coplanar workspace to provide highly conformal treatments of irregularly shaped tumours, resulting in an enhanced ability to spare surrounding healthy tissue and structures in close proximity to the treated volume.

It is capable of isocentric or non-isocentric treatment delivery to a targeting accuracy of 0.5 mm for targets unaffected by motion and 0.7 mm to those affected by motion, for example in the case of tumours affected by respiration.

Treatments are image guided using two kV x-ray beams projected from the treatment room ceiling (see x-ray sources, Figure C.110), which pass through the patient onto imaging tablets imbedded in the floor (see image detectors, Figure C.110). These images are compared to gold standard, digitally reconstructed images taken from the treatment planning CT; any movement in the position of the target or patient can then be measured and automatically corrected for by repositioning of the beam geometry to ensure submillimetre accuracy throughout treatment.

A typical treatment is usually delivered in between 1 and 5 fractions (6–25 Gy); treatment time varies from 30 to 90 min (depending on the location, shape, size and complexity of the tumour).

Respiratory-induced motion of tumours, for example in the case of lung, can cause significant targeting uncertainty. With Cyberknife, the motion of fiducial markers placed inside the tumour prior to treatment is correlated with that of external LED optical markers placed on the patient’s chest or abdomen.
During treatment, the surface markers can be tracked with an optical camera (see Optical imaging system, Figure C.110), allowing the machine head to move and follow the motion of the tumour in real time when the correlation between the two is known. The x-ray imaging system captures the position of the internal fiducial markers at intervals during treatment so that the correlation can be updated and the motion of the treatment head corrected if necessary.

**Cyclotron (Nuclear Medicine)**

A cyclotron is a charged particle accelerator commonly used to produce radionuclides for nuclear medicine imaging and radiotherapy. Many institutions and hospitals have their own cyclotron for onsite production of short-lived radionuclides. The basic principle is to accelerate electrically charged particles, typically protons, and direct them towards a target. Accelerated particles can cause a nuclear reaction in the target content and produce radionuclides. The typical cyclotron design is described in Figure C.111 and it consists of two opposite semicircular metal electrodes, often referred to as 'dees' because of their shape. The two dees are positioned horizontally between the two poles of an electromagnet. In the centre between the two dees is an ion source, S, used to generate charged particles. This entire ‘package’ is kept in vacuum at \( \sim 10^{-3} \) Pa.

During operation, the particles are accelerated in the electric field in the gap between the two dees. Inside the dees, there is no electric field, just a magnetic field that will curve the particle beam. The proton beam will follow a circular path until extraction of the accelerated beam.

The particle energy in a cyclotron system is given by

\[
E(\text{MeV}) = \frac{4.8 \times 10^{-3} (H \times R \times Z)^2}{A}
\]

where

- \( R \) is the radius of the particle orbit in cm
- \( H \) is magnetic field strength in tesla
- \( A \) and \( Z \) are the atomic number and the mass number of the accelerated particle, respectively

Consequently, the principal limitations in particle acceleration are dees radius and the magnetic field strength. When the particles are accelerated to the maximum energy, they can be extracted in two different ways; a target can be placed in the particle beam path, which is called internal beam irradiation. The other approach is to extract the beam from the cyclotron and direct it towards an external target, i.e. external beam irradiation. The stripping (i.e. the extraction of the beam) method differs from positive and negative ion cyclotrons. Positive ion cyclotrons use an electrostatic deflector to divert the beam onto the target. A negative ion cyclotron typically accelerates \( \text{H}^- \) ions (one proton and two electrons). When the beam has reached its outermost orbit, it is passed through a carbon foil where the two electrons are ‘stripped’ from the atom, leaving only a proton and a positive charge. The proton will bend in the opposite direction onto the target. Using this stripping technique, one could extract two beams simultaneously by using two foils. The first foil is positioned so that only half of the beam is stripped, leaving the rest of the beam for the second target. Due to the unstable nature of the \( \text{H}^- \) ion, the negative ion cyclotron must be operated at \( 10^{-5} \) Pa (\( 10^{-3} \) for positive ion cyclotron).

**Related Articles:** Internal beam irradiation, External beam irradiation


**Cyclotron electrodes (dees) (Nuclear Medicine)**

Most cyclotrons consist of a pair of hollow semicircular metal electrodes shaped like a \( D \), hence the name. The dees produce a magnetic field that is used to bend the particle beam. They are housed in a vacuum chamber that is in a uniform magnetic field provided by an electromagnet. Between the dees, there is maintained a potential difference that alternates in time. This potential difference creates an electric field across the gap between the dees. An ion source is placed in-between the two dees to produce charged particles. The particles are accelerated by the electrical field in the space between the two dees and bent in the magnetic field inside the dees (note that the electrons are only accelerated in the gap between the dees) (Figure C.111).

**Related Article:** Cyclotron

Cyclotron target

(Nuclear Medicine) A cyclotron target is a chamber with a gas, fluid or a solid piece of material onto which a particle beam is directed to produce radionuclides. For $^{18}\text{F}$ production, the ideal target is constructed in a material with a few key characteristics:

1. Chemically inert
2. High thermal conductivity
3. Low activation of target chamber
4. Low reactivity of the $^{18}\text{F}$ fluoride produced
5. Low contaminants in the $^{18}\text{F}$ fluoride produced

The target body should be chemically inert to prevent the produced radionuclide from reacting with the target body. The high-intensity beam irradiates a small area of the target and produces a lot of thermal energy and it is therefore important to have an adequate cooling system and a material with good heat conduction, for example, in a water target with insufficient cooling, the water may evaporate due to the temperature rise. Additional desired features are low activation of the target body to prevent unnecessary radiation dose to cyclotron operators and staff, low reactivity of the $^{18}\text{F}$ produced and minimal contaminations to the final product.

Cyclotron-produced radionuclides

(Nuclear Medicine) Radionuclides produced in a cyclotron often have similar characteristics. Typically, positive particles are accelerated and added to the nucleus:

1. Therefore, most cyclotron products lie below the line of stability. These isotopes tend to decay via EC and $\beta^+\text{ decay.}$
2. The atomic number of the product isotope is often changed in the process, which changes the chemical properties. The product is therefore often considered to be carrier-free.
3. Since the cross section for protons is smaller relative to neutron cross section, the quantities of radioactivity produced in a cyclotron is lower than in a reactor. Therefore, the price on cyclotron products tends to be more expensive than reactor products.

Regardless of the price, there are a number of good reasons for cyclotron production. A cyclotron provides onsite production of short-lived radionuclides like $^{11}\text{C}\ (T_{1/2} = 20\text{ min}),\ ^{13}\text{N}\ (T_{1/2} = 10\text{ min}),\ ^{15}\text{O}\ (T_{1/2} = 2\text{ min})$ and $^{18}\text{F}\ (T_{1/2} = 110\text{ min}).\ ^{11}\text{C},\ ^{13}\text{N}$ and $^{15}\text{O}$ are common elements in biological substances and they can be labelled with a wide variety of biologically relevant tracers. $^{18}\text{F}$ is the most commonly used beta emitter and it is frequently labelled with a glucose analogue fluorodeoxyglucose (FDG) to study the glucose metabolism in the body. $^{18}\text{F}$ has a longer half-life, which allows transports from a regional cyclotron to adjacent hospitals. FDG is used primarily in oncology but also for brain and heart scans.

Cylindrical ionisation chamber

(Radiation Protection) Figure C.112 shows a schematic diagram of a cylindrical ionisation chamber filled with a gas. Cylindrical ionisation chambers are used for the determination of the radiation exposure (dose), for example a pocket dosimeter for x-rays. They are also applied for measuring absorbed dose in beams of heavy particles in depth to which the measured dose refers.

In the cylindrical ionisation chamber, the electric field lines are in radial direction and the strength $E$ of the electric field depends on the distance $r$ according to the following expression:

$$E = \frac{V}{(r \ln(b/a))}$$

where

- $V$ is a voltage applied to the electrode
- $r$ is a distance from the anode (inner electrode)
- $b$ is a radius of an outer electrode (cathode)
- $a$ is a radius of the anode

The average energy to produce an ion pair in air is $\sim 35\text{ eV}$ and for other typical gases is in range $20–40\text{ eV}.$ If a beta particle of energy about $1\text{ MeV}$ enters the chamber, it will be absorbed in the gas and as result the number $N$ of ion pairs produced, for example in air is

$$N = \frac{1\text{ MeV}}{35\text{ eV}} \approx 2.9 \times 10^4 \text{ ion pairs}$$

and it corresponds to an electric charge $Q = N \times e$, where $e = 1.6 \times 10^{-19}\text{ C}$, then $Q = 4.6 \times 10^{-13}\text{ C}$

The amplitude of the voltage pulse $V$ depends on the capacitance $C$ of the chamber as follows:

$$V = \frac{Q}{C}$$

The unit of radiation exposure is defined as the amount of electric charge produced in a unit mass of the air by x- or gamma radiation. It is expressed in the SI as $\text{C/kg}$ or in roentgen (R) used as traditional unit:

$$1\text{ R} = 2.58 \times 10^{-4}\text{ C/kg}$$

Related Article: Ionisation chamber

DAC (digital-to-analogue converter)
(General) See Digital-to-analogue converter (DAC)

Damping
(Ultrasound) The axial resolution in diagnostic ultrasound imaging depends on the length of the pulses used. To produce short pulse duration it is important to reduce the ringing effect in the piezoelectric transducer element so that the pulse length is only a few cycles long. Therefore the natural resonance of the element must be damped. Ringing describes the internal reverberations in the transducer element caused by the large difference (20 times) in acoustic impedance between PZT and soft tissue. A damping, or backing, layer behind the PZT will reduce the ringing (damp the vibrations) and the properties of this damping material should ideally have an acoustic impedance equal to PZT and with good absorption characteristics. Mounting matching layers to ensure efficient transmission of sound into the body in front of the transducer will also help to reduce the ringing effect.

The pulse duration is one property used to define the damping efficiency of the transducer. The bandwidth and $Q$-factor are others. A short pulse give rise to a broad bandwidth and a low $Q$. $Q$ is defined as the ratio of the transducer centre frequency to its bandwidth (Figure D.1).

Related Articles: Attenuation, Absorption, Matching layer, Bandwidth, Axial resolution

Dark blood
(Magnetic Resonance) Dark blood or ‘black blood’ imaging is a general term for MRI techniques used to null signal from flowing blood to provide improved blood/tissue contrast.

Basic spin echo (SE) sequences demonstrate an inherent black blood characteristic. Signal in SE sequences is generated through excitation of a selected slice with a $90^\circ$ RF pulse followed a time TE/2 later with a $180^\circ$ refocusing pulse. Spins associated with blood flowing into or out of the selected slice in a time <TE/2 will not be excited by both RF pulses and do not contribute strongly to signal. As a result blood flow in SE images will appear dark. This holds only where flow is perpendicular to the slice and is sufficiently fast. Slow flow or flow in the plane of the slice will be exposed to both the 90° and 180° pulses and will contribute to signal.

In double inversion recovery dark blood imaging two 180° preparatory pulses are used to null flowing blood (see Figure D.2). The first inverting pulse is non selective and inverts all longitudinal magnetisation within the receptive volume of the RF transmit coil. The second pulse is selective and restores longitudinal magnetisation in the selected slice. The inverting pulse has no net effect on the tissue in the slice. Inflowing blood is however inverted, with its longitudinal magnetisation recovering towards zero with a $T_1$ time constant. A delay time TI is left prior to starting the imaging sequence proper or ‘host’ sequence. The delay time is chosen so that blood longitudinal magnetisation is zero when the imaging sequence starts and cannot contribute to signal. Double inversion pulses are typically used in conjunction with a fast SE or single shot (i.e. HASTE) SE sequence.

In triple inversion recovery imaging, a third inversion pulse is added to null both fat and blood – essentially an extension of principles of the STIR imaging technique.

Black blood imaging techniques are used extensively in cardiac imaging and imaging of the great vessels and carotids. Figure D.3 demonstrates the tissue/blood contrast achievable in a double inversion recovery image.

Dark current
(Diagnostic Radiology) Dark current is the current (a small constant amount) through a photodetector (such as photodiode, photomultiplier PMT, TV camera tube, etc.) in the absence of incident light (radiation).

The dark current consists of the following basic components:

- Photocurrent – generated by background radiation
- Saturation current – at the semiconductor junction

In cases of precise measurements, dark current should be taken into account upon calibration of the photosensitive devices (PMT, photodiodes, etc.). It is also considered as a source of noise.

To reduce the influence of dark current over the total signal, it is compensated either by subtracting it from the final signal or reducing to 0.

Darkroom
(Diagnostic Radiology) A darkroom is a room in which radiographic or photographic film can be handled and processed safely without being exposed to light. Most types of film are not sensitive to all wavelengths or colour of light. Therefore a darkroom can be illuminated with light that is not in the sensitive range of the film to permit human vision in the area. This illumination is provided with ‘safelights’. The colour or wavelength of the light is determined by...
the type of light source or filter. This should be selected to ensure
that the light does not extend into the sensitive spectral region of the
films that are being used.

Quality assurance procedures include inspecting and testing a
darkroom for light leaks from the exterior and to ensure the film is
not sensitive to the safelight being used.

Related Article: Film processing

FIGURE D.2 Double inversion recovery black blood imaging.

FIGURE D.3 Cardiac double inversion MRI image showing good con-
trast between myocardium and blood.

Data acquisition PET
(Nuclear Medicine) There are two main types of data acquisition
in dedicated positron emission tomography (PET) scanners, two-
and three-dimensional acquisitions. In most dedicated PET scan-
ners the detectors are placed in rings along the through-plane.

2D Data Acquisition Mode: In old scanners the rings were
separated with axial lead septa between each ring. This acquisi-
tion mode is called 2D data acquisition, and only photons parallel
to the plane of the detector ring are allowed to reach the detector.
This means that a coincidence can only be registered between two
detectors in the same ring. This mode is not very sensitive to either
random or scattered coincidences. Another method for acquiring
data is to allow registration of photons in an adjacent ring. If the
septa are shortened, more photons will be able to reach the adjacent
rings. This method is more sensitive, i.e. has a higher count rate for
each individual detector. A line of response (LOR) between two
adjacent detectors is known as a cross plane and the LOR will cross
the direct plane in the centre of the scanner. To further increase
the sensitivity more cross planes can be included. This acquisition
method allows a high sensitivity, but does also degrade the spatial
resolution in the axial direction. There is also a higher occurrence
of random and scatter coincidence in the high sensitive mode than
in the conventional 2D mode.

3D Data Acquisition Mode: Some of the annihilation photons
absorbed in the septa in a 2D mode could provide true coinci-
dences. In a 3D mode PET scanner there is no septum between the
detector rings, which means that coincidences can occur between a
detector and a detector in any of the opposite rings. Such a scanner
has a very high sensitivity compared to one in a 2D mode. The axial sensitivity peaks at the centre of the middle ring(s). It is therefore of great importance to place relevant structures in the centre of the axial field of view (FOV). 3D data cannot be divided into independent 2D slices and reconstructed using filtered back projection. Instead a three-dimensional reconstruction algorithm is used. These algorithms are more time consuming than the conventional 2D reconstruction. But since PET scanners operated in 3D mode demonstrate such high sensitivity, 3D mode is available in new commercial PET scanners. Due to its high sensitivity PET is well suited for dynamic studies. With dynamic studies it is possible to gain knowledge about tissue function by calculating blood flow or studying radiopharmaceutical delivery, accumulation and clearance.

Related Articles: PET, Field of view, Beta decay


Database

(General) Database is a collection of data, structured in a specific way and stored in the computer system. There are various medical imaging databases collecting and organising images of anatomical structures in norm and pathology.

Daughter nucleus

(Nuclear Medicine) See Bateman equations in parent–daughter decay

Daughter radionuclides

(Nuclear Medicine) In a decay chain the initial radioisotope, the parent, disintegrates to another radioactive isotope. The second radioisotope is called the daughter radioisotope or decay product. The daughter radioisotope will eventually decay to a third decay product, or grand-daughter radioisotope. This chain will continue until one of the decay products is stable.

In nuclear physics, a specific part of a decay chain can be of clinical and/or research interest (the parent radioisotope can be chosen until one of the decay products is stable. The daughter radionuclide in this example is Tc99m. Tc99m has some suitable characteristics for imaging, namely the photon energy (140 keV) and a high number of gamma rays per disintegration (close to 1).

The Bateman equations describe parent–daughter decay.

Related Articles: Parent radionuclide, Grand-daughter radionuclide, Bateman equations, Molybdenum breakthrough

DCT (discrete cosine transform)

(General) See Discrete cosine transform (DCT)

Dead man’s switch

(Diagnostic Radiology) Switch which interrupts when released. Typical dead man’s switch is the foot switch used in x-ray fluoroscopy, which terminates the exposure when the foot is removed from the pedal-switch.

Dead time

(Nuclear Medicine) Dead time is the duration after each detection event where the detector is unable to acquire new signals even if they do happen. When several events are recorded near-simultaneously, the image detector system is unable to separate all the events which leads to a signal pile up. During a period of time the electronics cannot discriminate any new pulses in order to accept new events. This time span is referred to as dead time, i.e. no events are acquired.

A common approach to deal with high count rate is to use different types of circuitry, like discriminators or buffers. The latter one ‘holds’ a signal and passes it on to the next circuitry when it is ready. Another approach is to limit the read out time of the PM-tube signal, where the signal is set to zero after a specified time. This technique causes a degradation of the intrinsic spatial resolution and the energy resolution but it can still prove useful to avoid dead time losses.

Related Articles: PET, SPECT, Pile up


Dead time

(Ultrasound) Dead time, dead-zone, or ring-down distance is defined as the distance from the transducer face to the closest target that can be resolved. The transmit pulse may obscure this first part of the image due to the high amplitude reverberations of the pulse within the lens covering the transducer face. These reverberations take time to ring down to the level of small echoes. Most modern transducers have virtually no dead-zone due to greatly improved matching layers and pulse forming techniques.

The dead time can be found using a phantom with wire targets close to the surface. Usually there is a group of targets that are equally spaced axially. If, for instance, the third target can be resolved and the spacing is 1 mm, the dead zone is ‘less than 3 mm’.

Deadtime losses

(Radiation Protection) The period of time when a detector is insensitive for measuring a second event after the first one is called the ‘dead time’. For Geiger–Müller counters, dead time are on the order of 100–500 μs and for NaI(Tl), as well as for semiconductors, dead times are in the range from 0.5 to 10 μs.

As a consequence of the dead time of a detector the number of measured counts is less than the true one. The counts lost during the dead time are called dead time losses. If the counting rate is low the probability of an event arriving during the dead time is low. The dead time losses are a serious problem at high count rates and particularly for a paralysable counting system in which each event introduces its own dead time.

If the dead time is τd and measured count rate Nm then the true count rate Nt as equal to

\[ N_t = \frac{N_m}{1 - N_m \tau_d} \]

At low counting rates (Nm ≪ 1/τd)Nt = Nm.

At high rates for paralysable counting system: Nt = Nm exp(−N_m τ_d).

EXAMPLE:

If τ_d = 500 μs for a GM counter then for \( N_m = 10 \text{ cp} \) (10 ≪ 2 × 10^5): N_t ≈ N_m but for \( N_m = 10,000 \text{ cp} \) (10,000 < 2 × 10^5) \( N_t < N_m \) the correction must be made since \( N_m \tau_d \) is about 5.

Related Articles: Non-paralysable counting system, Paralysable counting system


Decay
(Radiation Protection) See Radioactive decay

Decay constant
(Nuclear Medicine) The decay constant \( \lambda \) of a radionuclide in a particular energy state is the quotient of \( dP \) by \( dt \), where \( dP \) is the probability that a given nucleus undergoes a spontaneous nuclear transformation from an energy state:

\[
\frac{dP}{dt} = \lambda
\]

In radioactive decay, the constant \( \lambda \) relates the rate of decay \( (dN/dt) \) of radioactive atoms to the number of radioactive atoms \( (N) \):

\[
\frac{dN}{dt} = -\lambda N
\]

The constant \( \lambda \) will have units of s\(^{-1}\), h\(^{-1}\), day\(^{-1}\) or other inverse time unit. Other equivalent names for the constant are transformation constant and disintegration constant.

For any given radioactive atom there is a constant probability that it will decay in a stated period, therefore the number of atoms remaining will decrease in an exponential fashion:

\[
N = N_0 e^{-\lambda t}
\]

This equation is the solution of the preceding equation and states that the number of radioactive atoms \( (N) \) at time \( t \), is equal to the initial number, \( N_0 \), multiplied by the factor \( e^{-\lambda t} \).

The real lifetime of any particular radionuclide ranges from 0 to \( \infty \); however, for a large number of parent nuclei the mean lifetime, \( \tau \), of a radioactive parent substance equals the sum of lifetimes of all the individual atoms divided by the initial number of radioactive nuclei in the radioactive substance at time \( t = 0 \). The decay constant \( \lambda \) and the mean life \( \tau \) are related through

\[
\tau = \frac{1}{\lambda}
\]

It is more common to use the half-life, \( T_{1/2} \), of a radionuclide, which is the mean time taken for the radionuclides in the particular energy state to decrease to one half of their initial number. The relation between \( \lambda \) and \( T_{1/2} \) is

\[
T_{1/2} = \frac{\ln 2}{\lambda} = \frac{0.693}{\lambda}
\]

Related Articles: Activity, Half life, Radioactive decay


Decay factor
(General) The term 'decay factor' is a general term used in connection with the function \( y = a \cdot b^x \). When the value of \( b \) is less than 1, \( b \) is sometimes called the decay factor as 1 is then dealing with exponential decay. Where \( b \) is positive, \( b \) is termed the growth factor.

The most common application of the equation \( y = a \cdot b^x \) in medical physics is in radioactive decay:

\[
N = N_0 e^{-\lambda t}
\]

where

- The number of radioactive atoms \( (N) \) at time \( t \) is equal to the initial number
- \( N_0 \) is multiplied by the factor \( e^{-\lambda t} \)
- \( \lambda \) is most often called the decay constant but other terms used are transformation constant and disintegration constant

Related Article: Radioactive decay

Decay scheme
(General) A decay scheme, in the context of radioactivity, is a two-dimensional representation of the changes and transitions involved in the transformation of a radionuclide into a stable nuclide or a different radionuclide. The axes represent energy (ordinate) and the proton number (abscissa). Decay schemes can be complicated and care must be taken to check whether the diagram has been simplified. Usually any neutrinos emitted are omitted.

A decay scheme will normally show
- Initial radionuclide: Symbol, with atomic mass and atomic number, and half-life.
- Mode(s) of decay: Particle(s) emitted, with percentage or mean number.
- Nuclear energy levels involved of daughter nucleus and gamma rays emitted.
- Daughter nuclide: Symbol, with atomic mass and atomic number.

However, due to complexities of many decay schemes, some of the information may be placed in an accompanying table. Figure D.4 and Table D.1 shows some of the information about the decay of \(^{131}\)I to \(^{131}\)Xenon. As beta decay is involved the \(^{131}\)Xenon is to the right compared to the \(^{131}\)I. It should be noted that not all transitions lead to gamma rays and internal conversion occurs.

Related Articles: Beta decay, Decay series, Gamma ray, Gamma radiation, Radioactivity, Radionuclide


![Figure D.4 Decay of \(^{131}\)I. Transitions and energy levels involving <1% disintegrations have been omitted.](image-url)
 Decay series  
(Radiation Protection) A decay series is when there is a series of nuclear transformations, starting with a radionuclide (‘the parent’) and eventually ending in a stable nuclide. The parent nuclide decays into a different nuclide (‘the daughter’) which is itself radioactive and which decays in turn to yet another daughter and the series continues until the final daughter is a stable nuclide.

Most naturally occurring radionuclides are members of one of three series, named the uranium, actinium and thorium series. The final stable nuclide of each series is one of the stable isotopes of lead.

Related Article: Radioactive decay

Decibel (dB)  
(Ultrasound) In ultrasound contexts the decibel notation is used for comparing pulse pressure amplitudes, intensities and for expressing attenuation. The definitions are

Relative pressure amplitude level (dB) = 20 \log\left(\frac{p_2}{p_1}\right)

Relative intensity level (dB) = 10 \log\left(\frac{I_2}{I_1}\right)

Note that: 10 \log\left(\frac{I_2}{I_1}\right) = 10 \log\left(\frac{p_2}{p_1}\right)^2 = 20 \log\left(\frac{p_2}{p_1}\right)

Some useful amplitude and intensity ratios expressed in dB are shown in the table.

<table>
<thead>
<tr>
<th>dB</th>
<th>\frac{p_2}{p_1}</th>
<th>\frac{I_2}{I_1}</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>3</td>
<td>1.414</td>
<td>2</td>
</tr>
<tr>
<td>6</td>
<td>2</td>
<td>4</td>
</tr>
<tr>
<td>20</td>
<td>10</td>
<td>100</td>
</tr>
<tr>
<td>40</td>
<td>100</td>
<td>10,000</td>
</tr>
<tr>
<td>60</td>
<td>1,000</td>
<td>1,000,000</td>
</tr>
</tbody>
</table>

Related Articles: Intensity, Attenuation

Decommissioning  
(Radiation Protection) Decommissioning is defined as the processes carried out such that an item of radiation equipment may be taken out of clinical use, or such that a radiation facility may safely be handed over to an alternative use, whether by the employer, or by another organisation.

In areas where unsealed radioactive sources have been used this may involve de-contaminating the area of any residual contamination. In areas where sealed radioactive sources are used, this may involve removing the sources from any equipment which to be disposed.

Decontamination  
(Radiation Protection) Decontamination is the removal or reduction of contamination by a physical or chemical process.

Related Article: Contamination


Deconvolution  
(General) Deconvolution is a mathematical process to find the input function, which has previously undergone convolution with some other function. Formally, the problem can be expressed as follows:

\[ f \ast g = h \]

where

- \( h \) is the known output
- \( g \) is the convolution function
- \( f \) is the desired input function that is unknown
- \( g \) may be the impulse response function of a known system, in which case it can be fully deconvolved (in the absence of noise)

However in many cases \( g \) is not known and it needs to be estimated before deconvolution can be attempted.

The convolution theorem is a useful tool in deconvolution – to extract the input distribution, the Fourier transform of the output, \( \text{FT}(h) \), can be divided by the Fourier transform of the convolution function, \( \text{FT}(g) \), and then take the inverse Fourier transform:

\[ f = \text{FT}^{-1}\left(\frac{\text{FT}(h)}{\text{FT}(g)}\right) \]

This is straightforward for explicit functions however will not be precise if the functions are unknown or contain noise, since the Fourier transforms may not be fully defined.
In imaging, deconvolution is used to remove distortions or blurring that the input image may have undergone during the imaging process. The response of the imaging system (i.e. g) is known as the point spread function (PSF). Deconvolution is also an integral step in the filtered back projection algorithm for tomographic imaging.

**Abbreviations:** FT = Fourier transform and PSF = Point spread function.

**Related Articles:** Convolution integral, Point spread function, Filtered back projection

**Decoupling**

(Magnetic Resonance) Decoupling refers to the elimination of the interactions that give rise to spin coupling in systems of interacting spins.

The simplest approach to decoupling is intense, continuous RF irradiation at the resonance frequency of one of the coupled spins (say A). This has the effect of flipping spins rapidly between orientations (±½ in the case of spin ½ nuclei such as protons), so that the effect on the other spin species (say X) averages out.

Sometimes, however, decoupling of spins across a broader frequency range is required. In such instances, a variety of broadband decoupling schemes using composite pulse trains of increasing complexity are available: primarily WALTZ and its variants. These sequences rely on the fact that the effects of coupling within a heteronuclear AX spin system can be eliminated by inverting the A spins, and employ composite pulses designed to give good inversion over a wide frequency range.

Application of decoupling techniques on clinical MRI equipment requires special hardware modifications: an additional radio-frequency channel and coil (frequently a surface coil, although the body coil may be used). From a safety perspective, the additional power deposited by the decoupling channel must be taken into account when calculating specific absorption rate (SAR).


**Decubitus**

(General) There are a series of terms used to describe the position of an individual when undertaking different imaging examination.

Decubitus: Lying on the side. For example, imaging the right kidney in the left lateral decubitus position (Figure D.5).

**Related Article:** Patient position

**Deep inspiration breath hold (DIBH) technique**

(Radiotherapy) For treatment sites subject to the effects of breathing motion, it may be desirable to control the patient’s breathing in order to freeze this motion and reduce the severity of the breathing. One approach is the use of active breathing control (ABC – system supplied by Elekta). Another is to get the patient to hold his breath at full, or close to full, inspiration during irradiation in deep inspiration breath hold (DIBH). Often several breath holds are needed to cover the full treatment duration. DIBH may be particularly useful for tangential irradiation of the left breast, where it may reduce the volume of heart treated to high dose, or to exclude the heart from the high dose region completely.

**Abbreviations:** ABC = Active breathing control, also Active breathing coordinator and DIBH = Deep inspiration breath hold.

**Related Articles:** Active breathing control, Gating – respiratory, Image guided radiotherapy

**Deep therapy**

(Radiotherapy) Around the early 1970s the term ‘deep therapy’ was still given to treatments using energies of 200–300 kV. This was to distinguish it from ‘superficial therapy’ which was the term given to treatments using energies of around 60 to 140 kV. However with the coming of high energy beams, this energy range was called orthovoltage.

Today’s classifications would be as follows (see IPEMB, 1996):

<table>
<thead>
<tr>
<th>Energy Level</th>
<th>kV Range</th>
<th>mm Al</th>
<th>mm Cu EHV</th>
<th>mm Cu HVL</th>
</tr>
</thead>
<tbody>
<tr>
<td>Very low energy</td>
<td>8–50 kV</td>
<td>0.035–1.0</td>
<td>4</td>
<td>10–300</td>
</tr>
<tr>
<td>Low energy</td>
<td>50–160 kV</td>
<td>1.0–8</td>
<td></td>
<td>0.5–4.0</td>
</tr>
<tr>
<td>Medium energy</td>
<td>160–300 kV</td>
<td></td>
<td></td>
<td>0.5–4.0</td>
</tr>
<tr>
<td>Teletherapy/megavoltage therapy</td>
<td>Cobalt 60 and 2–25 MV linear accelerators</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>


**Defective pixel**

(Diagnostic Radiology) Defective pixel is a term used to access the status of pixels on LCD screens and/or CCD/CMOS sensors in digital cameras. Each pixel that does not perform as expected is considered defective.

Defective pixels functional classification:

1. Hot pixels – always on
2. Dead pixels – always off
3. Stuck pixels – one or more sub-pixels are always on or always off
4. New pixels – newly installed pixels

The number, type and location of defective pixels for each device/manufacturer are important criteria influencing the image quality and the functional status of the device.

**Related Article:** Bad pixel

**Deflection electrode**

(General) See Deflection plates in cathode ray tubes

**FIGURE D.5** Left lateral decubitus position.
Deflection plates in cathode ray tubes

Related Article: Delta rays

Delta function

(General) The delta function has two possible forms.

Kronecker Delta: The first is the discrete form, the Kronecker delta, \( \delta_j \), which is a single non-zero data value at the time indicated, and zero at other times:

\[
\delta_j = \begin{cases} 
1 & \text{if } i = j \\
0 & \text{if } i \neq j 
\end{cases}
\]

Often the Kronecker delta is simplified by setting \( j = 0 \). This form is often called the unit impulse:

\[
\delta_i = \begin{cases} 
1 & \text{if } i = 0 \\
0 & \text{if } i \neq 0 
\end{cases}
\]

Dirac Delta Function: Alternatively the continuous form is known as the Dirac delta function, and has the following properties:

\[
\delta(x) = \begin{cases} 
\infty & \text{if } x = 0 \\
0 & \text{if } x \neq 0 
\end{cases}
\]

\[
\int \delta(x)dx = 1
\]

This is not strictly a function but is often treated as such. It can be thought of as an infinitely sharp spike of infinite narrowness, with the value 0 everywhere except at \( x = 0 \), such that its integral area is unit. As with the Kronecker delta, it is often referred to as the unit impulse function.

The Dirac delta function has the following characteristic:

\[
\int f(x)\delta(x)dx = f(0)
\]

The Dirac comb consists of a pulse train of Dirac deltas, at uniform intervals, and is an often used sampling function used in signal processing.

The Fourier transform of the delta function is unity:

\[
\text{FT}[\delta(x)](k) = \int e^{2\pi ikx} \delta(x)dx = 1
\]

\[
\text{FT}[\delta(x-x_0)](k) = \int e^{2\pi ik(x-x_0)}dx = e^{-2\pi ikx_0}
\]

Related Article: Discrete Fourier transform

Delta rays

(Radiation Protection) When an energetic charged particle or photon passes through tissue it causes ionisation. Some ionising events may release an electron with sufficient energy to cause further ionisations. These secondary electrons may also be referred to as delta rays.

Related Articles: Secondary ionisation, Secondary electrons
Demagnification factor

(Diagnostic Radiology) The main components of an image intensifier are: input phosphor, photocathode, electron optics and output phosphor. The incident x-rays are absorbed in the input phosphor and light photons are emitted which interact with the photocathode to release photoelectrons. The photoelectrons produced in the photocathode are collected and focused by a series of electrostatic electrodes on the output phosphor, where thousands of light photons are emitted for each photoelectron collected.

The number of photoelectrons within the image intensifier does not change but, as the area of the output screen is smaller then the area of the photocathode, the number of electrons/mm² will increase.

The demagnification factor, also called minification gain, can be expressed as

$$\text{Demagnification factor} = \frac{(\text{Diameter of input phosphor})^2}{(\text{Diameter of output phosphor})^2}$$

Considering that the diameters of input phosphors range from 15 to 40 cm and a typical diameter of an output phosphor is 2.5 cm, the demagnification value is between 36 and 256.

Demagnification factor does not affect the fluoroscopic image quality, but it influences the image brightness. In fact, the brightness gain of an image intensifier is obtained as the product of demagnification factor and flux gain.

Related Article: Image intensifier

Demodulation

(Ultrasound) In general, demodulation refers to the extraction of an information-carrying signal from a high frequency signal (carrier). An example from radio electronics is amplitude modulation, where the amplitude of a high frequency signal is varied according to an audio signal.

In diagnostic ultrasound, there are two principal applications where demodulation is employed. The first is in greyscale imaging, where the echo from a surface or reflector has the shape of a pulse that oscillates for a wavelength. The information-carrying signal is actually only that there is a surface, which desired to be shown in an image as a thin line. This information is extracted by finding the envelope of the echodata (often referred to as RF-data, RF for radiofrequency). This can be done by first rectifying the signal, and then filter the result to obtain a smooth envelope.

The second is in Doppler systems, where the information-carrying signal is the Doppler shift, i.e. a frequency modulation of the carrier frequency. Here, the demodulation is performed as a multiplication of the received signal by the transmitted, so that both sum and difference frequencies are obtained. The sum frequencies are then filtered out, leaving the difference frequency – the Doppler shift. This is also usually done as a quadrature demodulation, where the demodulation frequencies is 90° out of phase by the other. This gives the possibility to obtain both positive and negative Doppler shifts.

Densitometer

(Diagnostic Radiology) A densitometer (Figure D.7) is an instrument used to measure the optical density or opacity of a translucent type film such as used for film/screen radiography. The optical density (OD) is determined by passing a small beam of light through the film area and measuring the amount of light passing through. From the ratio of the light value penetrating the film (I) to the value of the light source without the film in place (I₀) (representing a density value of zero) the actual density is calculated and displayed:

$$\text{OD} = -\log \left( \frac{I}{I_0} \right)$$

Density correction

(Radiotherapy) The electron energy loss is theoretically the same for the same mass of material while in reality the stopping power is smaller for a condensed medium due to the polarisation of atoms. The polarisation effect influences the soft collision that is an energy transfer between the charged particle and the relative distant atoms. In gases the atoms are spaced widely and therefore they undergo interactions independently of each other. On the contrary in a condensed medium the density is increased by a factor of about 10³–10⁴ compared to a shortened distance between atoms of about 1/10 of the distance between atoms in a gas. The mass collision stopping power is therefore decreased in condensed media because of dipole distortion of the atoms near the track of the passing particle that weakens the coulomb force experienced by more distant atoms. Sternheimer et al. (1982) determined a term δ to be subtracted from the mass collision stopping power for electrons and positrons to correct for the polarisation effect. The term δ is given by

$$\delta = \sum_i f_i \ln \left( \frac{(\nu_i + \iota)}{\nu_i - \iota} \right) - \iota(1 - \beta^2)$$

where

$$\beta = \frac{v}{c}$$

with v being the velocity of the passing particle

$$f_i$$

is the oscillator strength of the ith transition, whose frequency is $$\nu_i$$

$$\iota$$

is a dimensionless frequency pertaining to I which is the solution of the equation

$$\frac{1}{\beta^2} - 1 = \sum_i f_i \nu_i + \iota$$

Dental radiography
(Diagnostic Radiology) Radiography is used by the dental profession to evaluate teeth and associated conditions.

The small receptors are designed to fit within the mouth.

Dental x-ray equipment usually uses low power x-ray tubes and generators. The simplest ones use fixed kV and mA and just control the length of the exposure with a timer. The most complex ones — (ortho pan tomographic equipment, OPG) make a special radiograph of the whole jaws.

Dephase
(Magnetic Resonance) Dephasing is the transverse part of the relaxation process in which the spins lose their coherence and return to their original state, and become more and more invisible to the receive coil.

Dephasing is the cumulative loss of phase which always occurs as soon as the RF-pulse is turned off. It is the result of spin-spin interactions following the small differences in precessional frequencies always present in the volume of interest.

As it causes a signal loss it is often a problem solved by shortening the echo time or by using a SE sequence. It may also improve contrast however, e.g. in DSC-imaging or diffusion-imaging. The dephasing of spins can be increased by use of paramagnetic contrast agents.

Inhomogeneities in the magnetic field can cause unwanted dephasing which can be detracted with shimming gradients.

Related Articles: Relaxation, RF-pulse

Deposition of dose
(Radiotherapy) Ionising radiation deposits dose, i.e. absorbed energy per unit of mass by ionisations and excitations. The energy deposited by ionising radiation is localised in few atoms. Charged particles produce these effects through direct Coulomb-force interactions with orbital electrons of the matter being traversed while x-ray and gamma interact by processes in which a secondary charged particle, normally an electron, is generated in the absorber and the secondary particle in turn deposits most of the energy. Uncharged particles as neutrons deposit energy through secondary charged particles by various process as a function of the neutron energy.

Related Article: Stopping power

Depth dose curve
(Radiotherapy) See Percentage depth dose

Depth dose distribution
(Radiotherapy) The dose distribution is a representation of the variation of dose with position in any region of an irradiated object. The dose distribution is determined from physical measurements in a phantom using an ionisation chamber or other radiation detector that can be positioned at various locations in the radiation beams of interest. The resulting data is presented in tabular form for calculating the linac monitor unit or time settings to deliver the prescribed dose to a representative point in the target volume and for calculating the relative dose to other points of interest. The dose distribution in the structures of a patient for a given incoming radiation field is also calculated by treatment planning systems (TPS) by dose deposition algorithm to provide the radiotherapist with a physical description of the treatment. The geometrical distribution of energy deposited by the radiation can be displayed in three dimensions by means of a set of graphs, each of which is a two-dimensional representation of the distribution in a plane. Lines of equal dose (isodose lines) are drawn as a function of two coordinates. Their standard display mode is colour-coded isodose lines overlaid on the greyscale image (Figure D.8).

FIGURE D.8 (See colour insert.) Colour code isodose lines.

Often dose distribution planar graphs are used for treatment planning in place of a complete set which of course would show the distribution in all three dimensions. A planar graph may be contracted further to a graphic representation of relative dose vs. depth in water measured from the surface where the beam enters. If this graph describes the dose in the centre of the field it is called a central-axis depth dose distribution (Figure D.9).

Several kinds of further techniques are available to display both dose and anatomical information using 3D solid surfaces as well as greyscale images. A ‘colour wash’ display technique assigns colour values corresponding to the dose to each pixel painting the whole dose distribution in transparent bands of colour on the greyscale image. By changing the dose to colour assignment in the display look-up table (LUT) the colour wash can be used to interactively scan a band of colour (dose) over the image interactively (Figure D.10).

FIGURE D.9 Dose distribution along the beam axis (percentage depth dose) for selected energy beams.
Depth gain compensation

(Ultrasound) Depth, or time, gain compensation (DGC/TGC) is the equalisation of echoes from different depths to compensate for increased attenuation with depth so as to display a uniform level of echoes from similar tissues at different depths. Depth gain compensation is sometimes referred to as time gain compensation/swept gain.

The transmitted ultrasound pulse and reflected echoes are attenuated by tissue due to absorption, reflection, scattering and beam dispersion and aberration. The attenuation is dependent on the acoustic properties of the tissue and its geometry.

Ultrasound scanners are designed and programmed with automatic compensation for this attenuation with increased gain from deeper echoes. This assumes a uniform predictable attenuation based on an average for soft tissue in the body, typically around 0.5–0.7 dB/cm/MHz. This may not be appropriate for a particular investigation. Examples where attenuation is low occur when scanning through fluid, e.g. bladder, amniotic fluid, cysts or ascites. Here deep structures appear too bright if the gain is uncorrected (Figure D.12), a phenomenon known as post cystic enhancement.

Conversely, uncorrected scanning through more highly attenuating tissue leads to dark images at depth because of increased attenuation.

Most ultrasound scanners have a depth/time gain control to enable the operator to modify the gain compensation. Typically these are a series of sliders (Figure D.14) corresponding to different depths within the image allowing gain to be increased or decreased at specific levels (Figure D.13). Since deeper echoes correspond to echoes with a longer transit to receive time, the terms depth gain control and time gain control are both used to describe this process.

A diagrammatic representation of the change in gain may be shown on the image as a line down side of the image showing relative gain settings at different depths (Figures D.12 and D.13).

The DGC and overall gain both affect the gain of the image. A common mistake made by those starting ultrasound imaging is to have the gain high, with excessive bright echoes obscuring contrast detail. This is a particular problem if scanning is done in rooms with high ambient light.

Related Articles: Attenuation, Absorption

Depth ionisation curve

(Radiotherapy) If a Farmer chamber is placed at different depths in water or phantom material and irradiated the electrometer reading

3D dose distributions can also be displayed and compared to solid surface display of the target volume and critical normal structures (Figure D.11).

FIGURE D.10 (See colour insert.) Colour wash display of dose distribution.

FIGURE D.11 (See colour insert.) 3D dose distribution.

FIGURE D.12 DGC – 1. With little change to the DGC (line R) the weak echoes and artefacts deep in the bladder are amplified and there are high intensity echoes in the bladder wall deep to the fluid.

FIGURE D.13 DGC – 2. With the DGC modified to reduce gain at depth there is a more uniform image of the bladder.
Depth of interaction

The depth of interaction (DOI) is an effect in PET imaging where the spatial resolution is degraded for off-centre radioactive sources. The photons from an off-centre source will in most cases enter the detector at an oblique angle. A substantial thickness of scintillator material is needed to stop photons. A photon with an oblique angle of incidence can penetrate one or several scintillators and be registered in an interaction in an adjacent detector (see Figure D.15). As a result an accurate spatial localisation of the event is impossible and the spatial resolution decreases.

The magnitude of the problem is determined by the width and length of each detector element and the diameter of the scanner. The effect also increases with the distance from the centre of the scanner.

Related Articles: Annihilation, Annihilation coincidence detection, Centre of rotation, PET


Depth of maximum dose

The depth at which $d_{\text{max}}$ occurs is called the depth of maximum dose and is energy dependent. For photons higher beam energies produce more forward scattering of electron and longer scatter path lengths. The higher the beam energy, the deeper the in tissue or phantom that $d_{\text{max}}$ occurs. See Figures D.16 and D.17.

Related Articles: Percentage depth dose, Build up, Build up dose

**FIGURE D.16** Central axis dose build up and falloff for different kV and MV photon beam energies.

<table>
<thead>
<tr>
<th>Photon Source</th>
<th>Build up Depth (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cs-137 rays (0.66 MV)</td>
<td>0.12</td>
</tr>
<tr>
<td>Cobalt-60 rays (mean energy 1.25 MV)</td>
<td>0.5</td>
</tr>
<tr>
<td>4 MV x-rays</td>
<td>1.0</td>
</tr>
<tr>
<td>6 MV x-rays</td>
<td>1.5</td>
</tr>
<tr>
<td>8 MV x-rays</td>
<td>2.0</td>
</tr>
<tr>
<td>10 MV x-rays</td>
<td>2.5</td>
</tr>
<tr>
<td>16 MV x-rays</td>
<td>3.0</td>
</tr>
</tbody>
</table>

**FIGURE D.15** Depth of interaction effect. A photon that enters a scintillator at an oblique angle can penetrate one or more crystals and interact in an adjacent scintillator instead. As a result the line of response is misplaced.

**FIGURE D.17** A table of build up depths for common photon sources.
DESS (dual echo steady state)
(Magnetic Resonance) See Dual echo steady state (DESS)

Destructive interference
(Ultrasound) Two waves that travel together can, dependent on their respective phase, be observed as a wave that is the sum of the two waves' individual amplitudes (constructive interference), or add up to no apparent wave motion at all if the amplitudes are equal and the phases are opposite (destructive interference). In the figure two point sources emit continuous waves, and in certain directions the waves appear to be in phase, whereas in others, to be out of phase. As can be deduced from the figure, this depends on the difference in distance from the observation point to the respective sources. Distance differences that correspond to an integer number of wave lengths plus one half wavelength result in destructive interference (Figure D.18).

Detail resolution
(Diagnostic Radiology) Detail resolution, also known as visibility of detail, is the ability to form a visible image of small objects or structures, i.e. detail within an image. The major factor that limits detail resolution and visibility of detail is the blurring that occurs during the imaging process as illustrated in Figure D.19.

Detection efficiency
(Radiation Protection) Detection efficiency of a radiation measurement can be defined as absolute or intrinsic. The absolute efficiency $\varepsilon_{abs}$ is equal to the ratio of number of events registered by the counting system to the number of events (e.g. gamma radiation quanta, beta particles, etc.) emitted by the source:

$$\varepsilon_{abs} = \frac{\text{Number of events registered}}{\text{Number of events emitted by the source}}$$

The absolute efficiency depends on the geometry of measurement and on the detector efficiency called the intrinsic efficiency, $\varepsilon_{int}$.

The intrinsic efficiency, $\varepsilon_{int}$, is defined as the ratio of the number of registered events to the number of events (e.g. gamma radiation quanta, beta particles, etc.) incident on detector:

$$\varepsilon_{int} = \frac{\text{Number of registered events}}{\text{Number of events incident on detector}}$$

The geometrical efficiency $\varepsilon_g$ (acceptance) is equal to the ratio of the number of particles (the term particle can include photons in the case of electromagnetic radiation), striking the detector to the total number of particles emitted by the source:

$$\varepsilon_g = \frac{\text{Number of particles striking the detector}}{\text{Number of particles emitted by the source}}$$

The absolute detection efficiency can be expressed as the product of the intrinsic efficiency and the geometrical one:

$$\varepsilon_{abs} = \varepsilon_{int} \times \varepsilon_g$$

Usually the detection efficiencies are expressed in percent.
Detective quantum efficiency (DQE)

The Geiger–Müller counter intrinsic efficiency for beta radiation is 100% and few per cent for gamma radiation. Liquid scintillation counting system can be used for tritium (beta particles with energy of ∼18keV) with an absolute efficiency about 100%.

**Related Articles:** Geiger–Müller counters, Liquid scintillation (LS) counting


**Detective quantum efficiency (DQE)**

(*Diagnostic Radiology*) The detective quantum efficiency (DQE) of an imaging system is a measure of both the efficiency with which the imaging system detects photons and of the noise created in the image. It is used in diagnostic radiography as a standard image metric to quantify the quality of an imaging system. The DQE is always dimensionless, and can have a value no greater than unity. This can be used to infer, e.g. the minimum x-ray dose to a patient which will provide images of satisfactory quality in an x-ray radiograph. The DQE is defined (Equation D.1) as

\[
DQE = \frac{\text{NEQ}}{N}
\]

where \( \text{NEQ} \) is the noise equivalent quanta over the image area, being a measure of image quality, and \( N \) is the number of incident photons over this area. The NEQ of the image is derived from the square of the signal to noise ratio (SNR) measured in the image. It describes the number of quanta which would result in the measured SNR if the measurement system was perfect (i.e. only Poisson noise was present). For example, an imaging device may measure 100 photons over a unit area \( (N = 100) \). If this device was ideal, then the noise in this signal is defined as \( \sqrt{100} = 10 \) and the SNR is also equal to 10. If we are now to consider the same system but with a realistic detection efficiency of less than 1 (or one which introduces extra noise to the image), then the actual SNR which is measured could be say 5. The NEQ derived from this SNR would be \( 5^2 = 25 \) and the DQE would therefore be 0.25. This can be interpreted as the detector only being able to utilise 25% of the incident photons to create the image.

This discussion considers the image metrics DQE and NEQ to have no dependence on spatial frequency. For a real imaging detector however this is not the case and the aforementioned quantities will vary depending on the spatial frequency component in the image. The terms in Equation D.1 should therefore be replaced by their frequency-dependent counterparts \( DQE(f) \) and \( \text{NEQ}(f) \). Furthermore \( \text{NEQ}(f) \) must be described using the modulation transfer function \( \text{NEQ}(f) \) and the noise power spectrum \( \text{NPS}(f) \) (for further details see the article *Noise equivalent quanta (NEQ)*). The full form of the frequency-dependent DQE(f) can be written (Equation D.2) as

\[
DQE(f) = \frac{\text{NEQ}(f)}{q} = \frac{q \text{MTF}(f)^2}{\text{NPS}(f)}
\]

where \( q \) is the number of incident photons per unit area (the average uniform input). This has replaced the term \( N \) in Equation D.1 due to NPS being a measure of the noise power per unit area.

\( \text{MTF}(f) \) and \( \text{NPS}(f) \) are both linearised functions with respect to input intensity.

The interpretation of the frequency-dependent detector quantum efficiency (DQE(f)) is very similar to that for standard DQE. \( DQE(f) \) describes the efficiency of the detector but at particular spatial frequency component. The DQE is sometimes expressed by the squared output SNR and the squared input SNR as (Equation D.3)

\[
DQE = \frac{\text{SNR}_{\text{out}}^2}{\text{SNR}_{\text{in}}^2}
\]

but this form must be used with caution, e.g. it is only correct, if \( \text{SNR}_{\text{in}} \) corresponds to the ideal photon-counting detector rather than any other type of detector such as energy-integrating detector.

In 2003, the International Electrotechnical Commission (IEC) published an international standard for the measurement of the DQE. The methodology states that the measurement must be under certain standardised conditions for the x-ray energy spectrum and the imaging system geometry. Under such conditions the MTF can then be determined through the edge response technique and the NPS is measured through Fourier analysis of a flat field image (see *Modulation transfer function (MTF)* and *Noise power spectrum (NPS)* articles for further details). The incident photon signal \( q \) is determined through measurements of the incident air kerma at the detector face using an appropriate radiation meter (ionisation chamber).

**Related Articles:** Modulation transfer function (MTF), Noise power spectrum (NPS), NEQ (noise equivalent quanta), Signal to noise ratio (SNR)


**Detective quantum efficiency (DQE)**

(*Nuclear Medicine*) The DQE is defined as the relative number of quanta of incident radiation detected by a detector. DQE is a measure of the information conversion quality. Later DQE has been generalised for all detecting devices and the new definition is

\[
DQE = \left( \frac{\text{SNR}_{\text{out}}}{\text{SNR}_{\text{in}}} \right)^2
\]

where \( \text{SNR}_{\text{out}} \) and \( \text{SNR}_{\text{in}} \) are the SNRs of the detector out and in signals, respectively. The DQE can be considered as the product of all the signal or data conversion steps in an imaging system. For example, in a conventional scintillation camera the DQE includes the photocathode quantum efficiency.


**Related Articles:** Quantum efficiency, Conversion efficiency of photocathodes

**Detector**

(*Diagnostic Radiology*) A general term used either to describe an instrument used to measure/monitor radiation (e.g. germanium detector), or to observe radiation (e.g. image detector). The latter
also measures the radiation, but transforms it into a visual image. Image detectors are also known as image receptors.

In x-ray diagnostic radiology incident radiation is recorded and used to create a radiographic image of a patient’s anatomy. The major types of image detectors used in x-ray diagnostic radiology and their classification are shown in Figure D.21.

Traditionally, analogue film/screen systems were used for medical imaging. In recent years, analogue detectors are now being replaced by modern digital ones. Digital detectors include computed radiography (CR) which uses an imaging phosphor plate that fits into the original film/screen bucky and direct digital radiography (DDR) which usually takes the form of an integrated flat panel detector. Within the DDR class of detectors the image is formed by either indirect or direct x-ray conversion, which is illustrated in Figure D.22. In indirect conversion the incident x-rays photons are converted to visible light photons before detection, an example of which is the silicon diode detector. In direct conversion systems the incident x-ray photons are measured directly without the need for conversion, an example of this technology is the amorphous selenium flat panel detectors.

The classification of detectors used in diagnostic radiology can sometimes be misleading. As CR technologies were developed before DDR detectors some sources use the term DDR when referring to CR and vice versa. Also, the term direct detectors is used to refer to all direct digital detectors and just direct x-ray conversion detectors. Direct digital detectors may also be called integrated detectors. Likewise, CR may be called non-integrated detectors.

**Detector fill factor**

(Diagnostic Radiology) The detector fill factor, fill-in factor or fill factor is the ratio of the active area of a detector (area where signal can be detected) over the total detector area (active area + TFT switch) – see diagram in the article Flat panel detector.

\[
\text{Fill factor} = \frac{\text{Active detector area}}{\text{Total detector area}}
\]

It is often used in diagnostic radiology to describe flat panel x-ray imaging systems as the detector fill factor is usually less than unity as a proportion of the active surface is covered with thin film transistors and other circuitry. A low fill factor can require a higher patient dose to produce the same signal strength as a similar detector with a higher fill factor as any incident radiation falling upon an inactive component does not contribute to the final image signal.

This can also be described as the pixel fill factor which is the ratio of the active area of one pixel over the total pixel area. Related Article: Flat panel detector

**Detector PET**

(Nuclear Medicine) In an early stage of PET imaging NaI(Tl) was used as a detector. NaI(Tl) has some disadvantages, namely a low density and low atomic number of the scintillator making it less efficient in stopping annihilation photons (511 keV). Today, because of this reason, most dedicated PET scanners use denser high-Z detector materials. These detectors are often arranged in rings around the patient. By using a ring shaped detector it is possible to simultaneously collect all projections.

**Block Detector:** The block detector was designed by Casey and Nutt in the mid 1980s. The construction of the block detector allows smaller detectors (higher resolution) and a reduction of photomultiplier tubes used (less expensive). A scintillator segment or block (BGO or LSO) is cut into an array of smaller elements. Each segment is read out using four PM-tubes as illustrated in Figure D.23. An opaque reflective material is placed in between the cuts and is used to avoid scintillation light escaping from one detector element to another. The depth of the cuts varies depending on the position of the element. The spatial position is determined by the signal in the four PM tubes assigned to the detector segment:

\[
X = \frac{(\text{PMT}_A + \text{PMT}_B) - (\text{PMT}_C + \text{PMT}_D)}{\text{PMT}_A + \text{PMT}_B + \text{PMT}_C + \text{PMT}_D} \quad \text{(D.4)}
\]

\[
Y = \frac{(\text{PMT}_A + \text{PMT}_C) - (\text{PMT}_B + \text{PMT}_D)}{\text{PMT}_A + \text{PMT}_B + \text{PMT}_C + \text{PMT}_D}
\]

\(\text{PMT}_A, \text{PMT}_B, \text{PMT}_C, \text{PMT}_D\) etc. are the signal strength of the individual PM tubes. \(X\) and \(Y\) are then used to determine in which sub-element the event occurred. A few modifications have been suggested to the original block detector design. One is to use bigger PM tubes so that each detector segment is monitored by a quartet of four different PM tubes. This method is called quadrant sharing and it reduces the number of PM tubes used by a factor four. This method has its
Disadvantages; one of them is higher dead time losses since each PM tube handles signals from several detector segments. Another modification is to use two different types of scintillator materials arranged in two separate layers (known as a phoswich) with one upper and one lower. Each layer has a different decay time, so by analysing the decay time in the pulse the event can be localised to either the upper or lower level. This method can reduce the DOI effects by a factor of two. These detectors, often a combination of LSO and GSO, are hard to manufacture and the general detector performance is slightly degraded compared to a pure LSO detector of the same dimensions.

**Abbreviations:** BGO = Bismuth germinate, GSO = Gadolinium oxyorthosilicate and LSO = Lutetium oxyorthosilicate:Ce.

**Related Articles:** PET, Beta decay


**Detector scatter event** (Nuclear Medicine) Detector scatter events refer to events in which a photon undergoes Compton scattering inside the detector. If the scattering occurs in the crystal, the scattered photon can either interact in a different location in the crystal or escape. In the latter case it is possible to discriminate events by analysing the pulse height, which is proportional to the amount of energy deposited. If the energy deposited is less than the lower threshold in the energy window, the event is discarded. But if the scattered photon interacts in a different part of the crystal the total amount of deposited energy will be within the energy window. The detector will not be able to separate the two events and the registered event will therefore be placed somewhere in-between the two interactions. The same effect occurs at high count rates if two events are simultaneously detected. Misplacing detected events causes a loss of image contrast.

![Image formation steps used in x-ray detectors.](#)

**FIGURE D.22** Image formation steps used in x-ray detectors.

![Diagram of block detector.](#)

**FIGURE D.23** Diagram of block detector.
Another type of detector scatter event is a collimator scattered event, i.e. photons that scatter in the collimator before being detected. Such events can be discriminated using an energy window. If however the system energy resolution is low, some of these events might be registered as true events.

Related Article: Collimators


Deterministic effects
(Radiation Protection) The detrimental biological effects of exposure to ionising radiation seen at higher doses/dose rates. These effects occur above a threshold in all persons exposed and the severity increases with the dose received.

Related Articles: Non-stochastic effects, Stochastic effects

Deuterium
(General) Deuterium is an isotope of hydrogen with one proton and one neutron in the nucleus. It is stable and is neither radioactive nor toxic. Its natural abundance is 0.0115%. It is represented by either the chemical symbol D or 2H. Approximately one part in 5000 of the hydrogen in seawater is deuterium. The nucleus of deuterium is referred to as the deuteron.

Applications and Uses: Deuterium is used as a solvent in proton nuclear magnetic resonance spectroscopy and as a non-radioactive tracer to study chemical and biochemical processes, where its presence is measured by mass spectroscopy. It is also used in the form of heavy water (D₂O) as a moderator in nuclear reactors.

Related Articles: Hydrogen, Isotope, Neutron, Nuclear magnetic resonance, Proton

Developer
(Diagnostic Radiology) The developer is a mixture of chemicals that converts the invisible latent image in an exposed film to a visible image. The exposure activates some of the silver halide crystals which make them sensitive to the developer chemistry. During the development process the developer converts the activated silver halide crystals into small specks of black metallic silver.

Although there are some differences in the chemistry of developer solutions supplied by various manufacturers, most contain the same basic chemicals. Each chemical has a specific function in the development process.

Reducer: Chemical reduction of the exposed silver bromide grains is the process that converts them into visible metallic silver. This action is typically provided by two chemicals in the solution: phenidone and hydroquinone. Phenidone is the more active and primarily produces the mid to lower portion of the greyscale. Hydroquinone produces the very dense, or dark, areas in an image.

Activator: The primary function of the activator, typically sodium carbonate, is to soften and swell the emulsion so that the reducers can reach the exposed grains.

Restraint: Potassium bromide is generally used as a restrainer. Its function is to moderate the rate of development.

Preservative: Sodium sulphite, a typical preservative, helps protect the reducing agents from oxidation because of their contact with air. It also reacts with oxidation products to reduce their activity.

Hardener: Glutaraldehyde is used as a hardener to retard the swelling of the emulsion. This is necessary in automatic processors in which the film is transported by a system of rollers.

Related Article: Film processing

Development time
(Diagnostic Radiology) The development time is the duration that a film is actually in the developer chemistry. Since development is a progressive process, the development time is a factor that determines the amount or degree of development that is produced. If the development time is too short, some of the exposed silver halide crystals will not be converted and the density or darkness of the film will be reduced. If the development time is too long, some of the unexposed crystals will be converted resulting in excessive film density or darkness (fogging).

The optimum development time is determined by the design of the film, composition of the developer chemistry, and the developer temperature. In most automatic film processors it is set to a fixed value.

Related Article: Film processing

Dewar
(General) A Dewar or Dewar flask is a container designed to thermally insulate the contents and used to store or transport liquids at temperatures different to the surroundings.

Named after its inventor Sir Thomas Dewar, a Scottish scientist, and first patented in United States under the trade name ‘Thermos’, the Dewar or Thermos consists of two thin walled vessels separated by a vacuum to prevent the conduction and convection of heat, and with a reflective layer to reduce thermal radiation.

Originally made from glass with cork stoppers, some are now made from plastics and come inside integral plastic or metal container to provide a more practical and rugged device (Figures D.24 and D.25).

Large Dewar flasks are used for storing materials for long periods at the temperature of liquid nitrogen and are sometimes referred to as ‘cryostats’.

DFT (discrete Fourier transform)
(General) See Discrete Fourier transform (DFT)

Diagnostic radiology
(Diagnostic Radiology) Diagnostic radiology is the use of radiation for diagnosing, evaluating, and managing diseases and injuries.
Most diagnostic radiology methods produce images. X-radiation was the first and is still the most common radiation used in diagnostic radiology but radiation from radioactive sources are also used along with RF signals (in magnetic resonance imaging) and ultrasound.

Diagnostic radiology is also the medical specialty for physicians with the education and training to make diagnosis from the images and to conduct related medical procedures (radiologists).

Currently many countries use the term medical imaging as a broader professional descriptor than diagnostic radiology.

**Diagnostic x-ray**

(Diagnostic Radiology) Diagnostic x-ray is the use of x-radiation to produce images for diagnostic purposes. This includes radiography, mammography, fluoroscopy, and computed tomography (CT).

Different x-ray spectra are used for the various procedures to optimise the contrast characteristics and exposure to the patient. The KV (in the units of kVp) is the primary factor used to adjust the spectrum. Typical values are shown on Figure D.26.

Relatively low values (24–32kV) are use for mammography to produce the required high contrast sensitivity to visualise soft tissue structures and small calcifications. At the other extreme, high KV values are used for chest imaging to provide good penetration through bones.

**Diamond detector**

(Radiation Protection) Diamond is a material with a very large band gap (∼5.6 eV), and can be operated as a simple conduction counter by applying ohmic contacts to opposite faces of the crystal. The dosimeter is based on a natural diamond crystal sealed in a polystyrene housing with a bias applied through thin golden contacts. The interactions of ionising radiation induce a temporary change in the electrical conductivity of the diamond through the production of electrons and positive holes that have sufficient energy to be free to move through the crystal.

It may be used for detection of ionising radiation in an active mode as well as in a passive one. The atomic number of carbon (Z = 6) is close to soft tissue (Z = 7.5). The interactions of ionising radiation induce a temporary change in the electrical conductivity of the diamond through the production of electrons and positive holes that have sufficient energy to be free to move through the crystal.

The x-ray or gamma radiation undergoing a photoelectric interaction in diamond disappears and liberated electrons are free to move from the valence to the conduction band. Diamond detector working in active mode consists of natural diamond crystal sealed in a polystyrene housing (or a thin diamond polycrystalline membrane, about 100 μm) placed between two electrodes, e.g. Cr/Au. This detector connected to an electrometer may be used to x-rays and alpha particles dose measurement in the range up to 10⁶ Gy. Besides, the dose detection can be performed using diamond resistor. The diamond electrical resistivity is inverse proportional to the dose rate from x-rays, gamma radiation and electrons beam.

Diamond detector (Figure D.27) works also in a passive thermoluminescence mode.

The thermoluminescence is the emission of optical radiation in the form of prompt fluorescence when the diamond is heated. Diamond contains deep traps in the bandgap between the valence and conduction bands. Some of the electrons fall into the traps which are too deep to release them. If the energy is delivered to the diamond by heating it, the electrons can move up to conduction band and hence produce photon of visible radiation. The radiosensitivity of a synthetic diamond crystal depends on the conditions in which these crystals are grown. The diamond dosimeter must have sufficient signal reproducibility, low fading (loss of trapped electrons at the room temperature) and linear proportionality between signal and dose over a relatively wide range of doses as well as mechanical and chemical stability. The dose response can be represented either by the measured light intensity as a function of detector temperature in a glow curve or by the linearity factor f(D)
Diaphragm (collimator)

(Diagnostic Radiology) Diaphragm is usually only a metal piece (absorber) with an opening, mounted beneath the x-ray tube. It simply restricts the x-ray beam filed. Complex diaphragms are called collimators. All these tools are beam restrictors. They have metal shutters (usually lead) which allow changing the size of the irradiating field. Most often these devices include also a light localiser which projects a light field with the same shape and size as the x-ray beam field. Such devices are also known as light beam diaphragm (LBD). A typical LBD for x-ray radiographic devices is shown on Figure D.28.

The movable shutters of the collimator (LBD) produce most often a rectangular field (to match the size of the x-ray film), also known as jaws. However shutters used in fluoroscopy produce circular field (to match the image field of the Image Intensifier) – see Figure D.29.

The light beam inside the LBD uses a system of mirrors to mimic the x-ray beam size (Figure D.30). The coincidence of the light field and the x-ray field has to be checked during quality control of x-ray equipment. This usually uses a simple tool with copper marks, according to which the light field is adjusted and further exposed (Figure D.31). Displacement of more than 1 cm usually requires re-adjustment of the LBD mechanism.

Related Articles: Beam restrictor, Light localiser

DICOM-RT

(Diagnostic Radiology) Diaphragm is usually only a metal piece (absorber) with an opening, mounted beneath the x-ray tube. It simply restricts the x-ray beam filed. Complex diaphragms are called collimators. All these tools are beam restrictors. They have metal shutters (usually lead) which allow changing the size of the irradiating field. Most often these devices include also a light localiser which projects a light field with the same shape and size as the x-ray beam field. Such devices are also known as light beam diaphragm (LBD). A typical LBD for x-ray radiographic devices is shown on Figure D.28.

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Related Articles: Beam restrictor, Light localiser

DICOM-RT

DICOM-RT is a subset of the DICOM 3 standard applicable in the radiotherapy domain. It usually contains the images (e.g. CT slices) along with associated data such as patient

The diamond detectors are applied for the doses exceeding 1 Gy. Diamond detectors are attractive in megavoltage photon and electron beam measurements either for the high spatial resolution because of their small size or for their near tissue equivalence. (\(Z = 6\) compared to \(Z_{\text{eff}} = 7.42\) for soft tissue). Other advantages include high sensitivity, low leakage current and high resistance to radiation damage. Diamond dosimeters are designed to measure relative dose distributions in high energy photon and electron beams, especially in high dose gradient regions.

Due to the high density of carbon in comparison to air, the recombination of charge carriers is much more important for diamond detectors than it is for ionisation chambers. The recombination rate in a crystal, when an equilibrium number of free electrons is established, is proportional to square root of the rate of ion-pair production and hence to the dose rate. This is due to the increase in the probability of recombination with the number of vacant hole. The charge collection efficiency therefore decreases with the dose rate. If impurities are present, metastable states are introduced which trap many electrons which would otherwise recombine with holes. If the number of electron in traps is large, the proportional increase in the number of vacant holes can be almost independent of dose rate. This means that the recombination rate, and hence the efficiency of charge collection, is almost independent of the rate of ion-pair production, giving an almost linear increase in detector signal with dose rate.

The response of the diamond detector has been shown to be nearly independent of the incident photon energy, since the mass attenuation coefficient ratio of water to carbon is almost nearly constant.

The company General Electric introduced a similar detector using garnet (gemstone detector).
Dielectric

A dielectric material is a substance where charges are bound so that its conductivity is ideally zero. Even if a field applied to the dielectric produces no migration of charge, it can produce a polarization of the dielectric which consists in a displacement of the electrons with respect to their equilibrium positions. In a dielectric charges are not free to move far enough to completely cancel the effect of any external electric field but they can move far enough to cause a partial cancelation.

Related Article: Conductivity

Dielectric constant

The electric field within a dielectric is reduced by the factor $1/\varepsilon$ from that which would exist without the dielectric. $\varepsilon$ is called dielectric constant of the dielectric.

Related Article: Dielectric

Differential absorption

The concept of differential absorption describes any other situation where the component parts of an object have unequal radiation absorption (and scattering) characteristics. It is often associated with radiological image formation.

Radiological images are produced when the x-rays beam pass through an inhomogeneous object and reaches the image receptor system. When x-rays penetrate such inhomogeneous tissue, they are not homogeneously absorbed; some tissues absorb x-rays more efficiently than others.

The image receptor system displays these differences in absorption as variations in brightness across the image that describes the relative radio-opacity of each part of the body being imaged: e.g. bone is more radio-opaque than soft tissues, and therefore appears as a different brightness.

Differential cross section

The differential cross section is a description of the probability of scattering in a particular direction. In other words, the differential cross section can be defined by the fraction of particles that can be found within a given cone of observation of solid unit angle when a target is irradiated:

$$\frac{d\sigma}{d\Omega} = \left( \frac{\Phi_s}{\Phi_i} \right) \frac{S_u}{\Omega_u}$$

where
- $\sigma$ is the scattering cross section
- $\Omega$ is the solid angle
- $\Phi_i$ is the incident flux
- $\Phi_s$ is the scattered flux
- $S_u$ is the surface unit
- $\Omega_u$ is the solid angle unit
The integral cross section is the integral of the differential cross section on the whole sphere of observation:

\[ \sigma = \int \left( \frac{d\Omega}{d\Omega} \right) d\Omega \]

**Related Articles:** Radiation scattering, Cross section

**Differential scattering cross section**

(Ultrasound) The differential scattering cross section is defined as the time-averaged scattered power in the direction (\(\theta, \phi\)) per unit solid angle (the solid angle being related to the surface of a sphere in the same way as an ordinary angle is related to the circumference of a circle). In other words, this describes how sound is scattered with angular direction. Integrating this over the surface of a sphere with a radius larger than the particle yields the scattering cross section (hence the name differential scattering cross section). Of special importance is the differential scattering in the opposite direction of the incoming sound, called the differential back-scattering cross section.

**Related Articles:** Scattering cross section, Absorption cross section, Extinction cross section

**Differentiation**

(General) Differentiation is a mathematical method to obtain the rate of change or slope of a curve at a given point. Consider the function \(y = f(x)\). The slope \(s\) of the curve between the points \(x\) and \(x + \delta x\) is given by the equation:

\[ s = \frac{f(x + \delta x) - f(x)}{\delta x} \] (D.5)

As the value of \(\delta x\) tends to zero, the value \(s\) becomes the differential (the derivative) \(dy/dx\) of the function at the point \(x\):

\[ \frac{dy}{dx} = \lim_{\delta x \to 0} \left( \frac{f(x + \delta x) - f(x)}{\delta x} \right) \] (D.6)

If the function represents a straight line such as \(y = 2x + 3\), then the slope of the line is 2 and it is the same at every point. This can be demonstrated by putting these data into Equation D.6:

\[ \frac{dy}{dx} = \lim_{\delta x \to 0} \left( \frac{2x + 2\delta x + 3 - 2x - 3}{\delta x} \right) = 2 \] (D.7)

In nuclear medicine, differentiation is very evident in the law of radioactive decay. This states that in a sample containing \(N\) radioactive atoms of a particular radionuclide, the average decay rate \(\Delta N/\Delta t\) is given by the equation

\[ \frac{\Delta N}{\Delta t} = -\lambda N \] (D.8)

where \(\lambda\) is the decay constant and \(t\) is the time. In the limit as \(t\) tends to zero, this equation becomes

\[ \frac{dN}{dt} = -\lambda N \] (D.9)

Equation D.9 represents the decay rate or activity of the radioactive sample.

**Diffraction**

(Ultrasound) Diffraction is a phenomenon describing the effect on a wave that travels through some kind of an aperture. How the sound wave spreads out as it moves away from the aperture depend on the shape and size of the aperture and the wavelength. In ultrasound applications, the referred aperture is often the ultrasound transducer, Figure D.32, and in many cases the wanted beam should be as collimated as possible. The Huygen’s principle is frequently used to describe diffraction, side lobes and other wave phenomena. For example, the ultrasound distribution from a transducer surface can be calculated by dividing the surface area into a large number of point sources and add up the contributions from all of these. The waves from the fictive point sources interfere when in phase and cancel each other when out of phase. This theory opens up for the solution with many small transducer elements in a matrix to create a collimated beam, which is the case for modern B-mode imaging.

The diffraction pattern of a plane disc source can be divided into a complex near field (near zone or Fresnel zone) that has a cylindrical shape with approximately the same radius as the source, and the far field (far zone or Fraunhofer zone) that diverges with the angle \(\Theta = \arcsin (0.61\lambda/a)\) where \(a\) is the radius of the disc and \(\lambda\) is the wavelength, Figure D.33. The far field starts at the distance

**FIGURE D.32** The ultrasound field, Huygen’s wavelets. The near field and far field. (Graphs courtesy of EMIT project, www.emerald2.eu)

**FIGURE D.33** Pressure distribution from a circular disc ultrasound transducer. Near field close to the transducer surface to the left. (Graphs courtesy of EMIT project, www.emerald2.eu)
a^2/\lambda), from the source. This implies that a large disc compared to the used wavelength produce a long near field with less divergence angle. The pressure distribution from a circular disc transducer is shown in Figure D.34.

**Related Articles:** Transducer, Ultrasonic field, Side lobes

**Diffusion** *(Magnetic Resonance)* Diffusion is the phenomenon by which molecules or particles randomly migrate due to their thermal energy. This random motion is called Brownian motion and is not only dependent on the temperature, but also on the size and surroundings of the molecules. Free isotropic diffusion, which occurs in, e.g. a glass of water, is described by a Gaussian distribution with mean zero and a variance given by Einstein’s equation as

\[ \langle r^2 \rangle = 2nDt \]

where
- \( r \) is the displacement
- \( n \) is the dimension of the diffusion (\( n = 1 \) in diffusion MRI)
- \( D \) is the diffusion coefficient
- \( t \) is the time during which the diffusion is measured

In MRI, the self-diffusion of water is studied, where spin displacements produce signal decay due to two subsequent magnetic field gradients of opposite polarity (see Diffusion imaging).

**Diffusion encoding** *(Magnetic Resonance)* A spin group exposed to an RF pulse will exhibit a transverse magnetisation component. When exposed to a magnetic field gradient, this component will acquire an additional phase angle in the rotating frame of reference. A ‘particle’ moving in a magnetic field gradient will subsequently be affected by different magnitudes of the gradient field, depending on its trajectory. By submitting a particle in motion to two gradient pulses with the same magnitude but with opposite polarity (obtained by the 180° pulse), a net phase shift is therefore obtained (Figure D.35). The particle in black, which is seen to the left in Figure D.35a, is moving to a new position, which is indicated in the right image. The particle will be affected by gradients of different polarity and different magnitude (Figure D.35b). The first gradient will impose a phase shift, \( \phi_1 \) (Figure D.35c, left) and the second gradient will impose a phase shift in the opposite direction, \( \phi_2 \) (Figure D.35c, right). When the particle is moving the net effect will be a phase shift different from zero (see Figure D.35d).

A particle that is stationary during both gradient one and two, represented by a dotted arrow in Figure D.35b, will have the net phase shift, \( \Phi \), equal to zero.

The net phase shift can be determined from

\[ \Phi(t) = \gamma \int G'(\tau) x(\tau) d\tau \]

where
- \( G'(\tau) \) is the time-dependent gradient strength
- \( x(\tau) \) is the displacement along the direction of the diffusion encoding gradient
- \( \gamma \) is the gyromagnetic ratio

The net magnetisation of a system exposed to the two gradient pulses will be lower if the particles are diffusing and this will lead to phase dispersion and a subsequent loss in signal, see Figure D.36. The sensitivity of the diffusion encoding is determined by the \( b \)-value (see \( b \)-value). With a higher \( b \)-value the phase dispersion will be more enhanced.

**Related Articles:** \( b \)-value, Diffusion

**FIGURE D.34** Pressure distribution from a circular disc ultrasound transducer. Near field close to the transducer surface to the left. Side lobes are clearly shown. (Graphs courtesy of EMIT project, www.emerald2.eu)

**FIGURE D.35** A particle subjected to diffusion will acquire a phase shift different from zero when it is affected by two gradient pulses of equal amplitude but with different effective polarity.

**FIGURE D.36** In one voxel there are many spins groups and the net effect after two diffusion encoding gradients will be a phase dispersion reducing the amplitude of the net magnetisation vector.
Diffusion imaging

(Magnetic Resonance) Diffusion imaging is by now a well-established technique within parametric (or, in a broad sense, functional) MRI. Several different types of images can be derived, such as apparent diffusion coefficient (ADC) maps and fractional anisotropy (FA) maps. The technique can also be used for diffusion tensor imaging (DTI) as well as for tractography.

The diffusion imaging method is based on an observed signal decay occurring due to the self-diffusion of water in the presence of two subsequent pulsed gradient of opposite polarity. The gradient pulses impart a location-dependent phase angle on the spins. Since the sign is reversed for the two gradient pulses, spins are refocused in the case of no motion, but introduce a phase dispersion in the case of moving spins. The phase of each spin is proportional to the distance traversed between the two pulses as well as the strength and duration of the gradient pulses (see b-value) and consequently, the signal decay is proportional to the average displacement of the spins in combination with the b-value.

In diffusion imaging, the basis is the $T_2$-weighted image together with a diffusion weighted (DW) image obtained with a specified diffusion sensitivity (b-value). Depending on the purpose of the examination, a protocol can consist of one $T_2$-weighted ($b=0$) image and several DW images obtained for several diffusion encoding directions. For clinical practise, the diffusion sensitivity is commonly set to $b=1000$ s/mm$^2$.

In the case of stroke diagnosis, it can be sufficient to acquire DW images in three orthogonal directions, but if the purpose is DTI, diffusion encoding must be performed in at least six non-collinear directions. To obtain FA maps and tractographies of higher quality a larger number of directions is preferred.

Related Articles: Apparent diffusion coefficient (ADC), b-values, Diffusion tensor imaging (DTI), Fractional anisotropy (FA), Tractography

Diffusion spectrum imaging (DSI)

(Magnetic Resonance) Diffusion spectrum imaging (DSI) was introduced as an attempt to solve the problems of crossing and kissing fibres in diffusion tensor tractography (DTT). The method is based on a 3D probability density distribution, which can be determined if diffusion encoding is performed in a large number of directions. A parallel to this technique is the so-called q-space imaging technique where a large number of differently diffusion sensitised images are acquired, but whereas the normal q-space imaging technique provides information regarding diffusion distances and possible restrictions, the DSI results in a 3D probability density distribution showing the likelihood for different diffusion directions.

Related Articles: Apparent diffusion coefficient (ADC), Fractional anisotropy (FA), Relative anisotropy (RA)

Diffusion tensor imaging (DTI)

(Magnetic Resonance) Diffusion tensor imaging is a technique based on diffusion weighted imaging (DWI) that enables a three-dimensional representation of the water self-diffusion. The self-diffusion of water in cerebral white matter (WM) is faster along than across the fibre bundles. By measuring the ADC in at least six non-collinear diffusion encoding directions, the diffusion tensor can be estimated. Three different parametric maps are often calculated from a DTI measurement: (1) the mean diffusivity map (MD) defined as the mean value of the eigenvalues of the diagonalised tensor (i.e. one third of the trace); (2) the degree of diffusion anisotropy, determined from the variance of three eigenvalues, is often described by the FA; and (3) the preferred directional diffusion, determined from the direction of the principle eigenvector, commonly represented by a colour coded map, where blue colour corresponds to a superior-inferior direction, red to a left-right direction and green to anterior-posterior direction (Figure D.37).

Related Article: Fractional anisotropy (FA)

Diffusion time

(Magnetic Resonance) In diffusion-weighted (DW) magnetic-resonance imaging (MRI), when using a conventional Stejskal–Tanner pulsed gradient sequence, the diffusion time ($T_D$) is defined as

$$T_D = \Delta - \frac{\delta}{3}$$
where

\[ \Delta \text{ is the time between the leading edges of the pulsed gradients} \]

\[ \delta \text{ is the duration of the pulsed gradient} \]

In this case, the ramp-up and ramp-down times of the diffusion encoding gradients are approximated to zero. The diffusion time relates to the mean square distance \( \langle r^2 \rangle \) traversed by the diffusing molecules as

\[ \langle r^2 \rangle = 2D\Delta \]

where \( D \) is the diffusion coefficient. This equation implies that if the diffusing molecules are restricted within a confinement, the observed value of \( D \) (i.e. the apparent diffusion coefficient, ADC) would be dependent on \( \Delta \). This has indeed been observed on excised tissue (Assaf, 1999), but not in healthy volunteers (Clark et al. 2000).

**Related Articles:** Apparent diffusion coefficient (ADC), Diffusion weighting


**Diffusion weighting**

*Diagnostic Radiology* Diffusion weighting (DW) is a technique used in both chemistry and medicine for probing diffusion to understand microstructural characteristics with the use of magnetic fields. By applying two additional magnetic field gradients to a SE pulse sequence, the measurement becomes sensitive to translational molecular motion, i.e. to self-diffusion. The first applied gradient gives the molecules a ‘phase label’, while the second gradient partly refocuses the phase of the molecules, depending on the net-movement of the molecules.

The strength of the diffusion encoding is normally defined by the \( b \)-value, defined as \( b = (\gamma \delta G)^2 T_D \), where \( \gamma \) is the gyromagnetic ratio, \( \delta \) and \( G \) the duration and amplitude of the diffusion encoding gradients and \( T_D \) is the diffusion time. A higher \( b \)-value gives higher diffusion sensitivity for molecular motions. In clinical routine, a \( b \)-value of approximately 1000 s/mm\(^2\) is commonly used to generate DW images. DW imaging is most of all used for differential diagnosis and assessment of ischemic stroke.

**Related Articles:** \( b \)-value, Diffusion time, Self diffusion, Spin echo

**Digital detector array**

*Diagnostic Radiology* A digital detector array can describe any flat panel x-ray detector used in digital diagnostic radiology. It refers to the array of elements used to detect the incident radiation and convert it to an electrical signal. This signal is then used to form the pixels in the final radiographic image.

In most modern flat panel x-ray detectors the matrix of detector elements is referred to as an *active matrix* because each element signal is read out row by row in an *active matrix read-out*.

Imaging linear scanning systems (or x-ray film scanners) have a linear array (row) of detectors.

**Related Articles:** Flat panel detector, Matrix array, TFT (thin film technology)

**Digital display**

*Diagnostic Radiology* In diagnostic radiology digital displays are used to observe radiological images. Within a hospital digital image information is stored via the picture archiving and communication system (PACS) and the digital display is the users’ interface to these stored medical images. Although older PACS systems may use analogue cathode ray tube displays (CRTs), at present newer systems use digital liquid crystal displays (LCDs).

Medical displays are classified as either primary displays, used for initial diagnosis of medical images, or secondary displays which are used for reviewing images alongside a radiologists report. Display devices are categorised by Spatial resolution, contrast resolution, screen size, greyscale bit depth and pixel defects classification. Many different bodies have published recommendations for medical displays, as an example the UK Royal College of Radiologists (2008) guidelines: screen resolution of \( \geq 1280 \times 1024 \) native pixel array; a screen size of between 42 and 50 cm; a maximum luminance of between 170 and 500 cd/m\(^2\); a luminance contrast ratio 250:1 to 500:1; and a greyscale bit depth of between 8 and 10 bit.
The American Association of Physicists in Medicine (AAPM) task group 18 has published guidelines for the assessment of display performance for medical imaging systems. These guidelines recommend that the responsibility of display quality control falls upon the medical physics staff within the hospital, and gives descriptions of the suggested tests performed. The areas tested include reflection, geometric distortion, luminance, the spatial and angular dependencies of luminance (or viewing angle), resolution, signal to noise ratio, glare, chromaticity and display artefacts.

**Related Articles:** Radiology information system (RIS), LCD (liquid crystal display), Viewing angle, Artefact, Image artefact, Picture archiving and communication system (PACS)


**Digital fluoroscopy**  
(*Diagnostic Radiology*) Digital fluoroscopy is the method used in almost all contemporary x-ray fluoroscopic systems. It digitised the image from the output of the Image Intensifier, either directly (through a CCD camera), or indirectly (through TV camera and ADC). The digitised image is then stored in the system RAM memory. The final mage resolution depends on the image intensifier active input screen and the matrix size of the memory (e.g., 4096 x 4096 pixels). This memory can perform last image hold, and sometimes forms of digital subtraction (see article on Digital subtraction angiography (DSA)). Recording of the digital image (digital fluorography, recorded on hard disk or other digital memory) can be made with rates from 6 to 60 frames per second and more. Systems with such fast memory replace conventional cine fluorographic systems. Digital fluoroscopy applies various methods of image processing.

**Related Article:** Digital subtraction angiography (DSA)

**Digital image**  
(*Diagnostic Radiology*) Digital image is the display of a two-dimensional array of binary data stored in a computer. Each single data in the array is named picture element, or pixel, and its value is correlated to the image representation in the output device (e.g. monitor, printer). Usually the digital image derives from a sampling process in which a continuous analogue signal is converted into a digital one.

There are two main parameters of the sampling process:

1. The sampling frequency (or sampling rate), which represents the interval between a sample and the next one. In the case of digital image this is defined as number of samples per distance unit.
2. The sampling bit depth, which represents the number of bits reserved for storing each sample, so that it is related to the maximum amplitude of the sampling data (dynamic range).

In digital image the sampling frequency determines the pixel dimension (spatial resolution) while the sampling depth influences the range of the pixel colour levels displayed (contrast).

In medical digital images the pixel level can represent different parameters (depending on the techniques). For example, greyscale levels (as in CT and MRI), false colour scale (as in ultrasound and nuclear medicine images), etc.

**Digital imaging and communication in medicine (DICOM)**  
(*Nuclear Medicine*) DICOM is an international standard for handling, storing, printing and transferring medical image data and patient information. The standard defines which file format and network communication protocol to use. Ideally with DICOM; workstations, printers, servers, etc. will be able to communicate with each other and medical imaging hardware using a joint archiving and communication system. DICOM data files are divided into data sets. Each data set in DICOM data has the patient information included as a part of the data set and not as a separate header file. Each dataset usually contains some attribute with pixel information, i.e. an image or a series of images and the patient information can therefore never be separated from the picture. This is to ensure that the images are always attributed to the correct patient.

**Related Article:** Picture archiving and communication systems (PACS)

**Digital mammography**  
(*Diagnostic Radiology*) Digital mammography is a specialised imaging technique for imaging breast tissue that uses a digital detector instead of the traditional screen-film systems. As noted in the article on Mammography the technique is mostly used for imaging breast cancer but is also used for other breast diseases. Mammography requires specialised imaging techniques because it requires high contrast to separate normal breast tissue from carcinoma. These two tissues have very similar linear attenuation coefficients. In addition it requires high resolution to visualise microcalifications, which are small calcium specks that are an early diagnostic sign of breast cancer.

Scientific studies have shown that digital systems outperform screen film in women who are pre-menopausal or have dense breasts.

Several types of digital detectors have been adapted to mammographic units. These include indirect detectors and selenium detectors. In addition specialised computed radiographic systems have been developed for mammography.

**High Contrast:** Two achieve high contrast the imaging is done at low kVp. However since the image processing that is available in digital systems allows an increase in contrast somewhat higher kVs are used. Most of the contrast is generated using the photoelectric effect. While the traditional molybdenum anode is still widely used in digital systems, tungsten anodes with filters such as silver with a higher k-edge are becoming popular. This type of x-ray tube has low output, so a 65 cm source-to-skin distance (SSD) is usually used.

**High Resolution:** In order to achieve high resolution a small focal spot is used. This is about 0.3 mm. Digital systems are not able to achieve the spatial resolution that a screen-film system can. A resolution of less than 8lp/mm is typical. However because of image processing and improved contrast, the micro-califications are well visualised.

**The Mammography Unit:** The unit has a short source to image distance (SID) because of the lower output. The x-ray tube is tilted to improve resolution. The tube is mounted so that the central ray passes through the chest wall edge of the breast instead of the centre of the receptor. This improves visualisation of all of the breast tissue. The unit has a grid of about 5/1. The unit is equipped with a breast compression system. Compression has a number of benefits:

- It immobilises the breast. This is important since exposures of greater than 2 s are not unusual.
- It spreads the tissues to improve diagnostic accuracy.
• It reduces breast thickness which reduces Compton scatter.
• It makes the breast thickness more uniform. This reduces the dynamic range of the x-ray signal and therefore allows for higher contrast film.

The units also have a system for magnification imaging which is done without a grid and a 0.15 mm focal spot.

Mammography units are so specialised that they have a variety of specialised quality assurance techniques which are necessary to achieve constancy of image quality. One of the earliest and most well-known was designed by the American College of Radiology. This has been adapted to the digital environment. Most countries have a mandatory quality assurance programme.

In addition to the image quality improvements of digital mammography another important aspect is the elimination of the many film processing problems that arise in the screen film environment.

Most experts believe that screening mammography is an important factor in reducing the mortality from breast cancer.

Related Article: Mammmography

Digital radiography

(Diagnostic Radiology) Digital radiography is a technique that replaces the conventional screen–film receptor or computed radiographic receptor with a digital receptor. It has advantages in both a practical and engineering level. Digital receptors can be used with both fixed and mobile radiographic units. From a practical point of view, the digital radiographic receptor improves both the efficiency and safety of a radiographic examination. With conventional screen film and computed radiographic techniques the technologist (radiographer) must leave the patient to process the images. This increases the time required for the examination and because the patient sometimes may be left alone, this can lead to unsafe practices. In the digital radiography environment the images are immediately available to the technologist within the imaging room. This improves both the efficiency and safety of the examination.

Two general classes of digital radiographic receptors are available – direct and indirect. In the direct type receptor the radiation interacts directly with the detector. Selenium detectors are an example of the direct type. In the indirect type the radiation interacts with a scintillator and the light from the scintillator is converted to an electrical signal in a TFT array. Digital radiographic detectors based on caesium iodide are an example of the indirect type. In most cases, the use of a digital radiographic detector requires replacement of the entire radiographic unit.

Digital radiographic units usually have a better signal-to-noise ratio than screen film systems or CR. The resolution lies between the II tube must be digitised but the image from the flat panel detector is stored in memory. The analogue image from the Image receptor (image Intensifier with TV tube or flat panel detector) is not used anymore. However mask film and inverted contrast film) is not used anymore. However conventional and computed radiographic techniques produce excellent digital subtraction images. As these images are predominantly used in angiography, the method is known as DSA.

Figure D.39 shows a block diagram of a typical DSA system. The images from the Image receptor (image Intensifier with TV tube or flat panel detector) are stored in memory. The analogue image from the II tube must be digitised but the image from the flat panel detector is acquired in a digital format. The mask image is stored in one of the matrix RAM memories (M1), and the image with contrast

Digital reconstructed radiographs (DRR)

(Radiotherapy) The planar simulation x-ray film provides a beam’s eye view (BEV) of the treatment portal but does not provide 3D information about anatomical structures. CT, on the other hand, provides anatomical information and target definition but does not allow a direct correlation with the treatment portals.

A digitally reconstructed radiograph DRR is the digital equivalent of a planar simulation x-ray film and can be reconstructed from a CT data set using virtual simulation software available on a CT simulator or a TPS. It represents a computed radiograph of a virtual patient generated from a CT data set representing the actual patient. Just like a conventional radiograph, the DRR accounts for the divergence of the beam.

The basic approach to producing a DRR involves several steps: choice of virtual source position; definition of image plane; ray tracing from virtual source to image plane; determination of the CT value for each volume element traversed by the ray line to generate an effective transmission value at each pixel on the image plane; summation of CT values along the ray line (line integration); and greyscale mapping. An extension of the DRR approach is the digitally composited radiograph (DCR), which provides an enhanced visualisation of bony landmarks and soft tissue structures. This is achieved by differentially weighting ranges of CT numbers that correspond to different tissues to be enhanced or suppressed in the resulting DCR images.

Abbreviation: DRR = Digital reconstructed radiograph.

Related Article: CT simulator


Digital subtraction angiography (DSA)

(Diagnostic Radiology) Digital subtraction angiography is an x-ray imaging procedure used to increase the visible contrast of blood vessels. The digital method uses the idea of the classical subtraction angiography, where two images of the same anatomical region are used – one without contrast media (called the mask), the other one with contrast media (called the contrast image). The subtraction is achieved when the second image is inverted (black-white) and is superimposed exactly over the mask. The resultant image contains only the image of the difference between the two images (the vessels filled with contrast), as all other anatomical structures remain almost invisible (the result of the superimposition of identical, but inverted structures – i.e. image subtraction). The fact that the anatomical structures (e.g. ribs or skull) have been removed from the image increases the visibility of the blood vessels (which would otherwise be superimposed with the image of the ribs or skull). Figure D.38 shows two subtracted images – brain blood vessels on right and kidney vessels on left. The image on the left is more complex (called functional subtraction image), as it shows with colour the time when the contrast media reaches specific anatomical region.

Classical subtraction (superimposition of the two x-ray films – mask film and inverted contrast film) is not used anymore. However digital fluoroscopic and radiographic x-ray systems produce excellent digital subtraction images. As these images are predominantly used in angiography, the method is known as DSA.

Figure D.39 shows a block diagram of a typical DSA system. The images from the Image receptor (image Intensifier with TV tube or flat panel detector) are stored in memory. The analogue image from the II tube must be digitised but the image from the flat panel detector is acquired in a digital format. The mask image is stored in one of the matrix RAM memories (M1), and the image with contrast
is stored in the other matrix RAM memory (M2). Each pixel from each memory should correspond to exactly the same anatomical region (that is why it is imperative that the patient be absolutely still during the examination). After this both images go to the subtractor, where the content of M2 is inverted and one image is subtracted from the other pixel by pixel. The resultant subtracted image can be processed by several methods to increase the contrast and improve the SNR. Finally the subtracted image passes through a digital to analogue converter (DAC) and is displayed on a monitor.

The subtraction process can use various algorithms – linear, quadratic, logarithmic, etc. The logarithmic subtraction produces images with better contrast, as it amplifies more structures with less contrast, than images with high contrast. These algorithms influence directly the SNR. However they change the pixel values and these can only bear relative information. This information can be used as an indicator of the contrast media bolus in different time frames (functional imaging). Additionally DSA allows other digital measurements as percentage of stenosis (narrowing of the blood vessels), cardiac ejection fraction, etc.

The most common artefact in DSA is movement artefact, as it leads to displacement of both images prior subtraction and consecutive misregistration.

**Related Articles:** Digital fluoroscopy, Mask mode fluoroscopy


**Digital-to-analogue converter (DAC)**

*(General)* A digital-to-analogue device converts a digital signal to an analogue signal. Digital-to-analogue is abbreviated as DAC. The digital signal (usually binary) with discrete signal levels is converted to a continuous analogue signal that represents a physical quantity, e.g. a current, voltage or electric charge. The reverse conversion is performed using an analogue-to-digital converter.

DACs can be found in modern music players where the signal must be converted from a digital form, e.g. mp3 to an analogue signal in order to be heard through a speaker.

**Related Articles:** Analogue-to-digital converter, Analogue signal

**Digital video disc (DVD)**

*(Nuclear Medicine)* A digital versatile disk (DVD) is an optical disc storage media format. Compared to a regular compact disc (CD), a DVD can store more than six times the data. A DVD-ROM is a read only DVD, DVD-R and DVD+R can be written once and function like a DVD-ROM. On a DVD-RAM, DVD-RW and DVD+RW information can be erased and written multiple times. DVDs are one of the most common storage media formats used today. Future formats include blu-ray discs which can store even more information than a DVD. The wavelength used to read a DVD is typically 650 nm (red light) whereas blu-ray uses 405 nm (blue light).

**Dilution quenching**

*(Radiation Protection)* Dilution quenching in liquid scintillation spectrometry is caused by the reduction of light emission...
resulting from the dilution of a scintillation solution with sample. The scintillation mixture is prepared by dissolving a primary fluor, such as PPO (2.5-diphenyloxazole), with a secondary fluor as solvent, e.g. toluene, xylene or phenylxylethane. The radioactive sample is added to it for counting. Liquid scintillation spectrometry is used mostly for beta radiation emitters, e.g. H-3 or C-14, with maximum energy of beta-particles, respectively about 18.6 and 159 keV. If the samples contain a single beta-isotope only then counting (total number of counts or count rate) and not spectrometric technique is applied for measuring radioactivity. In order to determine the absolute activity of a sample we need to know the counting efficiency ($C_\text{e}$) defined as the number of recorded counts ($N_r$) to the number of disintegrations ($N_t$) in the sample in time $t$:

$$C_\text{e} = \left( \frac{N_r}{N_t} \right) \times 100\%$$

The dilution changes the energy transfer from beta-particles to phosphor molecules, resulting in reduced light emission. This reduction is greater for lower energy beta-particles such as for H-3. To minimise this effect it is necessary to prepare an adequate dispersion of sample and scintillation liquid or to determine the counting efficiency using either an internal or external standard.


Diode

(General) Electronic component allowing electric current to flow in one direction only. Diodes could be either electron-vacuum tubes or semiconductor devices (most often used today).

Semiconductor diodes are made of silicon or germanium. When these materials are doped with specific impurities they produce n-type (with electrons as majority charge carriers) or p-type (with positive holes as majority charge carriers). The combination of n-type and p-type semiconductors creates a p-n junction. When negative voltage is applied to the n-type and positive to the p-type, the diode is forward biased and passes DC current true. If the voltage across the p-n junction is reversed (positive voltage to the n-type and negative voltage to the p-type), a depletion layer is created between both semiconductors which stops the current flow (reversed biased). Diodes are the main components used in rectifiers. The x-ray high voltage generators use special power diodes which rectify the kV anode voltage (Figure D.40).

There are many types of diodes. For example, photodiodes are made from special light-sensitive semiconductors. These are used mainly in reversed bias and the light generates current through the depleted p-n junction. The current passing through this diode is proportional to the light intensity. Similarly, radiation can create electrical current through the depleted zone of a reverse biased diode (a type of photodiode) – an effect used in semiconductor dosimeters (Figure D.41).

**Diode detectors**

(Radiotherapy) The irradiation of a semiconductor material can result in the creation of a hole-electron pair, provided enough energy is given to the electron to raise it into the conduction band. Therefore a semiconductor detector is the solid state analogue of an ionisation chamber. Semiconductors can be used in the form of a silicon diode which is a p-n junction diode. The dosimeters are produced by taking n-type or p-type silicon and counter-doping the surface to produce the opposite type material. These diodes are referred to as n-Si or p-Si dosimeters, depending upon the base material. N-type or p-type diodes behave differently because their minority carriers are holes or electrons, respectively.

Due to its high density the sensitivity of a diode detector exceeds that of similar gaseous detectors by factor of several tens of thousands, which implies that a point-like detector can be designed with the sensitive volume of less than 1 mm³. In the boundary between two regions one of p-type and another of n-type silicon there is a depletion layer which is free of charge carriers. When the detector is operating with zero external voltage a potential difference of about 0.7 V exists over this depletion area, causing the charge carriers created by the radiation to be sweep away into the body of the crystal. The sensitivity of the detector depends on the lifetime of the charge carriers and consequently on the amount of recombination centres in the crystal, which is determined by the diode type, the doping level and the accumulated dose. The diode sensitivity decreases with the accumulated dose as the radiation induce recombination centres into the lattice.

The effect of radiation damage represents the main limitation of silicon diodes. Other effects related to the detector material have also to be considered. The output signal of the diode depends on the photon energy because of the higher atomic number of silicon ($Z = 14$) compared to soft tissue ($Z \sim 7$) and the resultant higher contribution to the signal from the photo-electric effect. The diode signal is also dose rate dependent because at high dose rates the recombination centres will be occupied resulting in a relatively lower rate of recombination. Moreover the radiation damage may change the dose rate dependence of diode.

An increase of the diode response with the temperature has been reported, as increasing the detector temperature causes the energy of the minority carriers to increase and their probability for escaping recombination also increases. The thickness and the shape of the build-up cap will influence the angular response of the diode.

Diodes can be used both for in vivo measurements and those inside phantoms. Diodes are particularly useful for measurements in high dose gradient areas such as the penumbra region and in small field dosimetry. They are also often used for measurements of depth doses in electron beams measuring the dose distribution directly in contrast to the ionisation distribution measured by ionisation chambers. Diodes are also widely used in routine in vivo dosimetry. To determine the diode calibration factor for in vivo dosimetry a set of correction factors has to be established to account for variations in diode response in situations different from the reference condition.
Dipolar coupling

(Magnetic Resonance) Dipolar coupling (also called dipolar-dipolar coupling or DD-coupling) refers to the direct magnetic interactions between nuclear spins. The effect is mutual and can arise both between spins in molecules and between spins in different molecules.

The dipolar coupling effect is utilised in MR spectroscopy, in which it gives information about the molecular structure (Figure D.42).

The term dipolar coupling is a general word, which can involve both indirect dipole–dipole coupling and direct dipole–dipole coupling. The indirect coupling, which also involves coupling to orbital electrons, is better known as J-coupling. This indirect coupling gives rise to multiple spectral peaks in MR spectroscopy.

Related Article: J-coupling


Dirac d-function

(General) See Delta function

Direct current (DC)

(General) Direct current (DC) is a form of electrical current which flows in only one direction through an electrical circuit and is typically constant in value.

Direct current is generated by batteries and can also be converted from AC electrical power which is the usual form of power supplied by the power companies.

DC electrical theory is simple, with conductors having only one property, their resistance, which relates the DC current flowing through them to the DC potential applied across them (Ohm’s Law):

\[
R = \frac{\text{Resistance (ohms)}}{\text{Current (amps)}} = \frac{\text{Potential (volts)}}{\text{Current (amps)}}
\]

Where the potential difference is measured in volts, and the current in amperes, the resistance of a conductor is given by

\[
R = \frac{V}{I}
\]

These units also define the power or electrical energy dissipated in conductors as

\[
\text{Power (in watts)} = \text{Potential (in volts)} \times \text{Current (in amps)}
\]

Related Articles: Alternating voltage, Alternating current, Direct voltage

Direct driven

(General) The term direct-drive refers to motors which apply their motion without the need for any gears, levers or pulleys, or any other indirect linkage.

Direct-drive electrical motors have the advantage of providing fast, accurate motion of the driven part, but usually at the disadvantage of increased cost of manufacture.

The great benefit of few moving parts means a longer lifetime with less servicing need, and the lack of hysteresis, elasticity and backlash in any ‘drive chain’ gives greater precision.

Where slow movement/high torque is required, a direct drive motor will typically be considerably larger.

Direct voltage (DC voltage)

(General) Direct voltage (DC voltage) is a measure of the difference in electrical potential between any two points, and assumes such potentials are constant. It is practically used in measuring the potential difference between points in electrical circuits, when, by using DC electrical theory, the DC through conductors, and the function of electrical circuits can be deduced.

Direct voltage sources include batteries and ‘DC power supplies’ which convert AC electrical power from the domestic electrical supply into DC.

DC electrical theory only holds for CONSTANT on non-varying electrical potentials and currents. (A more complex AC theory is needed to take account of additional phenomena that can occur where potentials vary.)

In DC electrical theory, Ohm’s law states that DC current through a conductor is proportional to the DC potential across it; and that each conductor has only that one electrical property, the ratio of potential to current, its resistance:

\[
R = \frac{\text{Resistance (ohms)}}{\text{Current (amps)}} = \frac{\text{Potential (volts)}}{\text{Current (amps)}}
\]

Thevenin’s laws add to DC theory by noting that all the DC voltages measured around any closed pathway of conductors will add up to zero, whilst at any point in an electrical circuit, the sum of DC currents flowing into that point equal the sum of currents flowing out of it. These combine to make the analysis of electrical circuits practical.

Related Articles: Alternating voltage, Alternating current, Direct current

Dirty radionuclides

(Nuclear Medicine) Dirty radionuclides (also referred to as ‘dirty emitters’) are radionuclides which emit certain particles or gamma photons that are not suitable for imaging or radionuclide therapy.

One example of a dirty radionuclide is the positron emitter ⁸⁶Y. PET imaging with this radionuclide can be used to quantify the uptake of the therapy radionuclide ⁸⁶⁶Y. (⁸⁶⁶Y is a pure beta emitter and therefore cannot be used for quantitative imaging). ⁸⁶⁶Y is considered a dirty radionuclide because it produces 300% gamma photons per disintegration (β⁺ yield is 33%). Imaging the ⁸⁶⁶Y distribution using a PET scanner will give rise to a number of unwanted spurious coincidences between one of the annihilation photons
and one of the gamma photons. Spurious coincidences degrade the spatial resolution by misplacing the line of response (see separate article on Spurious coincidences).

Related Article: Spurious coincidences


Discharge current
(General) A discharge current is current caused by a release of stored electric energy or charge from a capacitor, battery or other components that store electrical energy.

Discrete cosine transform (DCT)
(General) The most common data analysis transforms are the Fourier transform and the Laplace transform. The discrete cosine transform (DCT) is closely related to the discrete Fourier transform (DFT), but only uses the cosines (the real part) as the basis functions.

Since it only uses the real cosines as a basis, it is designed for ‘real’ data only, with even symmetry, resulting in only ‘real’ values in the frequency domain. It is used especially for image compression (JPEG), where images are analysed for their cosine frequency content, and only data with significant amplitude are stored. This is a lossy process, but significant storage savings can be made without visually degrading the image.

Abbreviations: DCT = Discrete cosine transform and JPEG = Joint photographic experts group.

Related Articles: Discrete Fourier transform (DFT), Lossy

Discrete Fourier transform (DFT)
(General) The DFT of a series $x$ refers to its representation in the frequency domain. It is the discrete version of the continuous Fourier transform.

Its mathematical formulism is as follows: A discrete sequence of complex numbers $x_0, x_1, ..., x_{N-1}$ can be expressed in terms of a series of harmonics, size $a_k$

$$x_n = \frac{1}{N} \sum_{k=0}^{N-1} a_k e^{(2\pi i/N)kn}$$

where the harmonics $a_k$ (which represent the amplitude and phase of the different frequency components) can be computed as follows:

$$a_k = \sum_{n=0}^{N-1} x_n e^{-(2\pi i/N)kn}$$

This representation means that the original sampled signal is fully represented by a maximum of $N$ frequencies, i.e. their frequency response repeats indefinitely in frequency. This ambiguity leads to the aliasing that is often apparent in discrete sampling.

The DFT has equivalent properties to the continuous Fourier transform – i.e. linearity, shift theorem, differentiation, integration and convolution.

The DFT can be computed efficiently using a FFT algorithm. FFT algorithms are often more efficient for discrete signals of length $2^N$, and zero-padding can be used to increase the signal length for greater efficiency.

Applications of the DFT include data compression and MRI image reconstruction ($k$-space).

For more information please see the following external link.

Abbreviations: DFT = Discrete Fourier transform and FFT = Fast Fourier transform.

Related Articles: Fourier transform, Fast Fourier transform, Lossy, $k$-space


Discrete wavelet transform
(General) The most common data analysis transforms are the Fourier transform and the Laplace transform. The discrete wavelet transform is another transform which uses wavelets as ‘base functions’.

A typical wavelet may look like a wave that starts with zero amplitude, and then grows through to a maximum and then ebbs away back to zero once again. The wavelets used in DWT are scaled and translated copies of a base wavelet (often denoted the daughter and mother wavelets, respectively).

Wavelets are bounded in time as well as frequency, and this means that they can allow a more accurate description of both frequency and time content, especially for non-periodic or non-stationary signals with discontinuities.

Related Article: Discrete Fourier transform

Discretisation
(General) Discretisation is the process involved when signals or images are sampled and converted into digital form.

In time varying signals, discretisation occurs in both time and amplitude – the signal first being sampled at regular intervals, and each sample then being ‘quantised’ by comparing the signal value against a fixed set of equal steps in amplitude of a reference – as found in an analogue-to-digital converter (ADC).

Images may likewise be discretised by using digital cameras or scanners, where both have multiple small optical sensors which each provide one quantised value representing the amplitude of the underlying image.

Abbreviation: ADC = Analogue-to-digital converter.

Related Article: Analogue to digital converter

Discriminator
(Nuclear Medicine) A discriminator is an electrical device which discriminates signals below or above a set threshold value. When the input signal exceeds the threshold, the discriminator generates an output pulse.

Disintegration, radioactive
(Radiation Protection) Disintegration is the transformation of one radionuclide into another by radioactive decay.

Related Articles: Radioactive decay, Radionuclide

Displacement
(Ultrasound) When an ultrasound wave propagates in a medium, the particles within the medium will start oscillating. The displacement is the particles distance, at a certain time, from its rest position. A transducer surface moving in a sine shaped movement will give rise to a sinusoidal pressure wave. The relations between displacement, particle velocity and pressure for such a wave (expressed in one dimension) are

$$\text{Displacement}: s = x_0 \cos(\omega t - kx)$$

$$\text{Particle velocity}: v = \frac{ds}{dt} = -\omega x_0 \sin(\omega t - kx)$$
Pressure: \( P = Zv = -Z\omega s_0 \sin(\omega t - kx) \)

If the pressure amplitude \( p \) is known, the displacement amplitude can be calculated as

\[
s_0 = \frac{p}{(ZW)} \quad \text{with} \quad p = 1 \text{MPa}, \quad Z = 6 \times 10^8 \text{Rayls}, \quad f = 10 \text{MHz}
\]

\[
s_0 = \frac{10^6}{(1.6 \times 10^8 \times 2\pi \times 10^6)} \approx 0.1 \mu \text{m}
\]

Related Article: Longitudinal wave

Distal

(General) Directional anatomical terms describe the relationship of structures relative to other structures or locations in the body.

Distal: Away from, farther from the origin (e.g. the hand is located at the distal end of the forearm)

Related Article: Anatomical relationships

Distance learning

(General) Distance learning is remotely obtained education. These days distance learning is mainly delivered through Internet (in this way transforming itself to e-learning). Among the many definitions for this teaching process is ‘e-learning – the delivery of content for this teaching process is ‘e-learning – the delivery of content (or grey) nuances by varying the number of ink drops per area (effectively creating a mix of the ink colours and the white background page). For example, an \( 8 \times 8 \) dither matrix in grey halftones will create 64 grey nuances (from 1 ink drop in the area – minimal grey, to 64 drops in the area – maximal grey).

The method is based on the limited resolution of our eyes (inability to distinguish closely spaced spots). The dithering uses special algorithms producing ink drops patterns (with four, six or eight different inks). The result is more colours of the printout (or grey nuances in B/W printing). Good photo printers produce little, if any, visible dither pattern in the light-coloured areas (where they are most noticeable).

In general this method reduces the spatial resolution (the dither area being related to the smallest visible detail). For example, a 1440 dpi printer using \( 8 \times 8 \) dither matrix will produce printout \( 1440/8 = 180 \) dpi (effective dots per inch), what will produce approx. 3.5 lp/mm resolution. The method is not anymore used in medical imaging.

Divergence

(Ultrasound) The diffraction pattern of a plane disc source can be divided into a complex near field (near zone or Fresnel zone) that has a cylindrical shape with approximately the same radius as the source, and the far field (far zone or Fraunhofer zone) that diverges with the angle \( \Theta = \arcsin(0.61\lambda/a) \) where \( a \) is the radius of the disc and \( \lambda \) is the wavelength, Figure D.31. The far field starts at the distance \( a^2/\lambda \) from the source. This implies that a large disc compared to the used wavelength produce a long near field with less divergence angle.

Outside the main lobe, at angles wider than \( \Theta \), side lobes can be found, Figure D.32.

Related Articles: Diffraction, Side lobes

Divergent beam edge

(Radiotherapy) Since the radiation used in radiotherapy originates from an effective focal spot the radiation field will diverge along the beam axis. The geometry of this is set by things like the size and the position of the collimators relative to the focal spot. The field size is determined at a reference distance from the focus, e.g. 100 cm focus to axis (centre of rotation or iso-centre) distance. It will diverge as the distance increases from the focus. The length or width \( (D) \) of the field will change as in Equation D.11 (see Figure D.43):

\[
D_{100}(x+100) = \frac{D_{100}(100+x)}{100} \quad (D.11)
\]

Diverging collimator

(Nuclear Medicine) Collimators with holes that diverge from the detector face are called diverging collimators. The diverging point
DNA targeting

for such collimators is typically located 40–50 cm behind the back of the collimator. These collimators project a minified non-inverted image on the detector. If the activity distribution extends outside the detector area, a regular parallel-hole collimator would fail to image the entire distribution in a single view. A converging collimator allows photons that originate from an event localised outside the area beneath the collimator to enter; hence a diverging collimator is useful to image an activity distribution greater than the field-of-view of the detector. The relationship between the image size, $I$, and object size, $O$, depends on the collimator front to convergence point distance $f$, the collimator to source distance $b$ and the collimator thickness $t$. The quotient between $I$ and $O$ is called the minification factor:

$$ I \approx \frac{f-t}{f+b} $$  \hspace{1cm} (D.12)

A conclusion drawn from this equation is that an increase in the collimator-to-source distance results in a larger minification. Differences in object depth result in differences in the minifications which may lead to severe image distortion.

The collimator’s contribution to the system resolution is

$$ R_{\text{coll}} = \left( \frac{d (l'_{\text{eff}} + b)}{l'_{\text{eff}}} \right) \times \frac{1}{\cos \theta} \times \left[ 1 + \left( \frac{l'_{\text{eff}}}{2f} \right)^2 \right] $$  \hspace{1cm} (D.13)

where

- $d$ is the hole diameter
- $l'_{\text{eff}}$ is the effective hole length, accounting for septal penetration
- $\theta$ is the angle between the central axis of the collimator and an off centre source as seen in Figure D.44.

The best spatial resolution is acquired when the source is located as close as possible to the collimator face.

The geometric efficiency of a diverging collimator can be expressed as follows:

$$ g = K \left( \frac{d}{l'_{\text{eff}}} \right)^2 \times \left[ \frac{d^2}{(d+t)^2} \right] \times \left[ \frac{f+l}{f+l+b} \right] $$  \hspace{1cm} (D.14)

where $K$ is a constant depending on the hole shape. The geometric efficiency is defined as the fraction of incident photons registered by the detector and it decreases with the distance from the face of the collimator. The collimator is useful in projecting large objects onto the detector but at the cost of spatial resolution and efficiency.

Related Articles: SPECT, Parallel-hole collimator, Diverging collimator, Converging collimator, Collimator, Collimator design, Collimator parameters


DNA targeting

(Nuclear Medicine) This term refers to the targeting of the cell DNA by certain radiopharmaceuticals. One example is $^{18}$F-FLT ([$^{18}$F]$^{3}$-fluoro-L-$^{3}$-deoxythymidine) that is an alternative to $^{18}$F-FDG when investigating the therapeutic response. The accumulation rate of $^{18}$F-FLT in the cell is proportional to DNA production, hence proportional to cell proliferation rate. A decrease in DNA production is one of the first responses to tumour therapy so $^{18}$F-FLT accumulation acts as a good indication of tumour response.

Related Articles: Tracer kinetic modelling, Receptor targeting, Antigen targeting, Neuroreceptor targeting, Glycolysis targeting, Apoptosis targeting, Hypoxia targeting


Doped

(Diagnostic Radiology) See Doping

Doping

(Diagnostic Radiology) Doping is the process of introducing impurities into pure materials. The crystal structure of doped materials contains defects which change the properties of the material. These defects are the reason for the formation of energy traps in the forbidden zone. Semiconductors of n-type and p-type are produced through the process of doping (e.g. boron is a p-type dopant and phosphorous is a n-type dopant). The sensitivity of luminescence detectors can also be improved by doping the pure material (e.g. NaI or CsI) with dopant such as Ti, Na, Tb, etc. The resulting crystals in this case will be NaI:Tl, CsI:Na, etc.
Doppler angle

(Ultrasound) The term Doppler angle is one used to describe the angle between the ultrasound beam and direction of flow in the vessel under examination. The Doppler frequency is proportional to the cosine of the Doppler angle, denoted by $\theta$ as shown in the Doppler equation:

$$f_d = \frac{2Vf_r\cos\theta}{c}$$

where
- $V$ is the flow velocity
- $f_r$ is the transducer frequency
- $c$ is the sound velocity in blood

The angle is also referred to as the beam/flow angle or angle of insonation. The lack of a defined term for this angle and the use of other angle terms in imaging, including beam steering angle in linear arrays, can lead to considerable confusion for those learning ultrasound. This is unfortunate in view of its fundamental importance in the production of good sonograms and for accurate measurement of velocities from the sonogram.

At Doppler angles approaching 90° there are poor quality sonograms as Doppler frequencies decrease. Low Doppler angles give higher Doppler frequencies and good colour and spectral Doppler images. With high Doppler angles there are increasing errors possible in the measured velocities as a result of operator errors and inherent errors in Doppler ultrasound. As a pragmatic limit to this, many operators choose not use Doppler angles above 60° so as to avoid large errors in measured velocity (Figure D.45).

**Related Articles:** Doppler equation, Pulsed wave Doppler

### Doppler effect

(Ultrasound) The Doppler effect is the change in frequency and wavelength caused by relative motion between a wave source and an observer/detector.

The effect of a moving observer or moving source is shown in Figures D.46 and D.47.

When an observer moves towards the source of a wave travelling through a medium at speed $c$, it meets the waves at an increased frequency by virtue of its own velocity $v$. Conversely, an observer moving away from the source detects a lower frequency. If the direction of the observer velocity is not directly towards the source then the received frequency is also dependent on the angle $\theta$ between velocity and the direction of the wave as it passes the observer.

In medical ultrasound, commercial Doppler devices use the pulse echo technique where the source and detector are stationary but the echo is scattered from a moving target. The ultrasound then undergoes a Doppler shift by virtue of the shift as sound insonates the scatterer with a further shift back from the moving scatterer. The effect is shown in Figure D.48.

This simplifies as follows to give the familiar ultrasound Doppler equation:

$$f_\text{d} = f_r \left( \frac{c + v \cos \theta}{c} \right) - \frac{c}{c - v \cos \theta}$$

$$f_\text{d} = f_r - f_s$$

$$f_\text{d} = f_r \left( \frac{c + v \cos \theta}{c} \right) - 1$$

$$f_\text{d} = \frac{c^2 + cv\cos \theta - c^2 + cv\cos \theta}{c^2 - cv\cos \theta}$$
Doppler equation

The following relation is ‘the Doppler equation’ in diagnostic ultrasound:

\[ \Delta f = \frac{2vf_0}{c_0\cos \theta} \] \hspace{1cm} (D.15)

where

- \( \Delta f \) is the Doppler shift frequency
- \( v \) is the velocity of the target
- \( f_0 \) is the transmitted frequency
- \( c_0 \) is the speed of sound in blood
- \( \theta \) is the angle between the transmitter/receiver combination and the velocity of the particle

It also assumes that the angle between the transmit and receive beams is small.

The entire derivation of the Doppler equation is somewhat lengthy to include here, but essentially the analysis can be divided into two parts. In diagnostic ultrasound, the Doppler effect is the result of a change in frequency due to sound being reflected off a moving target, the first part of the analysis is to observe the Doppler shift when the transmitter is stationary, and the reflector (receiver) is moving. In the second part of the analysis, sound is transmitted from a moving source to a stationary receiver. By combining these two cases and assuming that the target velocity is much smaller than the sound speed, the preceding Doppler equation is obtained. A full description with diagrams is given in the article Doppler effect.

Related Article: Doppler effect

Doppler imaging modes
(Ultrasound) The term Doppler imaging covers several different implementations of the basic technique including

- Continuous wave Doppler
- Pulsed wave Doppler
- Colour flow imaging
- Power Doppler

See Flow imaging

Related Articles: Continuous wave Doppler, Pulsed wave Doppler, Colour flow imaging, Power Doppler

Doppler phantom
(Ultrasound) A Doppler phantom is a test object to measure and assess the performance of ultrasound Doppler devices including continuous wave Doppler systems and colour flow and pulsed wave spectral Doppler in ultrasound scanners. Parameters that may be tested include Spectral Doppler:

- Penetration depth/sensitivity
- Velocity estimation accuracy
- Waveform index estimation accuracy
- Volume flow estimation accuracy
- Temporal resolution
- Intrinsic spectral broadening
- Accuracy of angle estimation

Colour flow main parameters:

- Lowest detectable velocity
- Highest detectable velocity
- Image spatial resolution and registration accuracy
- Temporal resolution
- Velocity resolution
- Tissue colour suppression performance

Test objects are broadly divided into

- Flow phantoms where a fluid is passed through flow channels within a medium with acoustic characteristics that mimic human tissue (Figure D.49)
- String phantoms where a moving band or thread is insonated (Figure D.50)

Doppler sample volume
(Ultrasound) The Doppler sample volume is the target volume from which pulsed wave Doppler echoes are obtained. The sample volume is determined by

- The axial length along the pulsed Doppler beam usually displayed on the image as a pair of parallel lines (see Figure D.47)
- The beam width of the Doppler beam at the selected depth
- The slice thickness of the Doppler beam, usually determined by the acoustic lens

In ultrasound scanners, the sample volume usually refers to the length along the beam over which the operator chooses to investigate flow. The length of the sample volume, also referred to as the range gate, can be varied in a series of specific distances, e.g. from

\[ f_d = \frac{2cv\cos \theta}{c - vc\cos \theta} \]

\[ v << c \]

\[ f_d = \frac{2f_0v\cos \theta}{c} \]
Doppler shift
(Ultrasound) See Doppler effect

Doppler ultrasound
(Ultrasound) The term Doppler ultrasound describes a range of techniques, instrumentation and devices to detect, image and measure blood flow by ultrasound.

It is conventionally described as follows: when ultrasound insonates moving targets, the movement causes the reflected echoes to change frequency. The change in frequency is dependent on the relative direction of ultrasound beam and blood flow, the velocity of the targets and the transmitted frequency as shown in the ultrasound Doppler equation:

\[ f_d = \frac{2Vf_t \cos \theta}{c} \]

where
- \( V \) is the velocity of the blood
- \( f_t \) is the transmit frequency
- \( \theta \) is the angle between flow and the beam
- \( c \) is the speed of sound in blood
- \( f_d \) is the resulting Doppler frequency

The velocity of blood in major arteries and veins in humans range from a few cm/s to around 6 m/s (in diseased arteries). The Doppler frequencies that result from diagnostic ultrasound transmitted frequencies are in the audio range. The time changing distribution of Doppler frequencies can be displayed as a sonogram and, if an estimate of beam/flow angle \( \theta \) is made, the sonogram shows the time varying distribution of velocities.

Two basic forms of Doppler ultrasound are commonly used clinically.

In continuous wave (CW) Doppler, a transducer consists of separate transmit and receive elements. The difference in frequencies is the Doppler frequency which is amplified, played through a loudspeaker (Figure D.51) and, in some systems, displayed as a sonogram (Figure D.52).

Pulsed wave Doppler (PWD) systems have the advantage that examination of flow at a specific depth can be made and PWD

FIGURE D.49 Sample volume of 1.5 mm in a flow phantom. The sonogram shows high constant velocities from the centre of the flow channel. The sample volume is visible as a pair of lines parallel to the transducer face in the centre of the vessel.

FIGURE D.50 Sample volume of 6 mm. The sample volume includes the entire vessel. The sonogram shows higher intensities generally and Doppler frequencies from the flow adjacent to the wall.

1.5 to 10 mm. This can alter the pulse length and the time over which the echo is processed for Doppler frequency calculation. Because pulsed Doppler techniques use narrowband processing, the minimum sample volume is significantly larger than the minimum B-mode axial resolution.

In operation, the sample volume is selected to suit the application. A small sample volume can be used to investigate flow in the centre of the vessel and allows for more precise investigation of vessels where there are several vessels in close proximity. For mean velocity measurements, the vessel should be uniformly insonated. The sample volume should then be enlarged to include the whole vessel, insonating flows in the centre of the vessel and at the vessel walls. However, altering the sample volume only alters the axial length; the slice thickness remains undetermined by the operator and in-plane focus, usually set to the sample volume depth, means the vessel under investigation may still not be insonated uniformly.

Related Articles: Pulsed wave Doppler, Laminar flow

FIGURE D.51 Hand-held continuous wave Doppler ultrasound device with different probes operating at different frequencies for specific applications. The unit gives an audio output of Doppler frequencies.
Dose

(Radiation Protection) This is an ambiguous term. It is normally used when referring to absorbed dose.

Related Article: Absorbed dose

Dose area histogram

(Radiotherapy) 2D or 3D histograms are used in radiotherapy treatment planning. 3D histograms, represented by dose volume histograms, are used more commonly due to their greater comprehensiveness. 2D histograms, represented by dose area histograms (DAH), provide 2D dose statistical information about the given 2D view (differential or cumulative/integral). DAH can show minimum, maximum and mean dose values for the selected structure.

Abbreviation: DAH = Dose area histogram.

Related Article: Dose volume histogram

Dose area product (DAP)

(Diagnostic Radiology) The dose area product is the most common method of quantifying patient dose in radiology.

It is the product of the air kerma at a distance from the source and the area that the x-ray beam covers measured at that same distance. Its units are typically cGy cm².

In practice the DAP is measured using a DAP meter, which consists of a parallel plate ionisation chamber, attached to the beam exit port of the x-ray tube, and a display, usually located near the control console (Figure D.53).

As the exposure is taken the ionisation chamber measures the air kerma of the beam. It then multiplies this by the area the parallel plates cover to give the dose area product, which is then displayed for the user. In this way DAP meters provide a simple, non-invasive and instantaneous method of assessing patient dose.

The simplicity of DAP measurements lie in the fact that the focal spot of the x-ray tube can be treated as a point source, so the air kerma will reduce as you move away from the tube in accordance with the inverse square law. At the same time the area is increasing with a square relationship, so the change to the dose and the area effectively cancels out, leaving the DAP measurement the same at all points along the beams path. Therefore the DAP at the patient is the same as that measured at the tube, by the DAP meter.

Related Articles: Stationary anode, Rotating anode, Target, Line focus principle, Biangular anode disk, Focal spot actual, Focal spot effective, Focal spot

Dose calculation

(Radiation Protection) This is the calculation of the radiation dose received by an individual. It may be the result of direct measurement, or indirect measurement with the use of conversion factors, or by the use of physical or mathematical phantom and Monte Carlo modelling of the absorption and scattering of ionising radiation travelling through the body.

Dose calculations are normally performed either before or after medical investigations involving exposure to ionising radiation (x-rays or nuclear medicine) and used to describe the risk to the patient, or to an unborn child of a pregnant patient.

Related Articles: Radiation exposure, Radiation dose, Organ dose, Effective dose, Equivalent dose

Dose calibrator

(Nuclear Medicine) A dose calibrator is used to measure the radioactivity in a sample usually a syringe used for injection in a patient or an experimental animal. The word dose calibrator is somewhat misleading and activity or radioactivity calibrator would be more appropriate. To measure the activity accurately, the device has to be calibrated. Efficiency curves must be derived to take into account the geometry of the sample (i.e. the volume) to get the efficacy in the measurement. A common dose calibrator consists of a well ionising chamber but can also consist of other detector materials.

Related Article: Activity measurement
Dose conversion factors
(Radiation Protection) Radiation dose is rarely measured directly. For example, an ionisation chamber measures exposure – the amount of ionisation caused by the radiation in a volume of air. The various dose quantities can then be derived from the measurement by the use of a number of different factors to convert from one quantity to another.

Related Articles: Exposure, Radiation weighting factors, Tissue weighting factors, Absorbed dose, Equivalent dose, Effective dose

Dose distribution
(Radiotherapy) Radiation interacts with matter and in so doing releases a dose at any point of the irradiated object. The representation of the variation of dose with position in any region of an irradiated object is called dose distribution. The dose distribution may be measured in a phantom or it may be calculated by a TPS. To measure accurately a dose distribution it is necessary to avoid that the detector disturbs the distribution and to realise a condition of charged particle equilibrium. The visualisation of a calculated dose distribution serves to optimise patient planning and to check the accuracy of the computer generated dose distribution.

Related Articles: Equilibrium dose distribution, Charged particle equilibrium

Dose equivalent
(Radiation Protection) Prior to the publication of ICRP 60 ‘1990 Recommendations of the International Commission on Radiological Protection’ published in 1991 Dose Equivalent was the parameter used to describe whole body effective dose due to a partial body irradiation. ICRP 60 renamed the parameter Equivalent Dose.

Related Article: Equivalent dose


Dose length product
(Diagnostic Radiology) The dose length product (DLP) is a dose descriptor used in CT scanning. It is an approximate indicator of stochastic radiation risk. The DLP is defined as the product of absorbed dose, as given by the volume CT dose index, CTDIvol, and the scan length – see the following equation and Figure D.54. It has units of milli-Gray centimetres:

\[
\text{DLP} = \text{CTDI}_{\text{vol}} \times L \text{ (mGy cm)}, \quad \text{where } L = \text{Scan length}
\]

Doubling the DLP in a given anatomical area will double the stochastic radiation risk. Generally, stochastic radiation risk is given by the effective dose, \( E \) (milliSieverts). Conversion factors are available to convert DLP to \( E \) (Table D.2).

Related Articles: Computed tomography dose index (CTDI), Effective dose

Further Reading: 2004 CT Quality Criteria (MSCT 2004), online from: www.msct.eu/CT_Quality_Criteria.htm

Dose limiting tissue
(Radiotherapy) Radiation treatment inevitably affects normal tissue, and the severity of these radiation induced effects may limit the dose that can be safely delivered to the tumour. Usually one healthy organ or tissue in the treatment field is more radiosensitive than the others; hence the patient cannot be exposed to levels of radiation higher than the tolerance of this dose-limiting tissue. For further details see the articles on Adverse effects, Therapeutic effect and Tolerance.

Related Articles: Adverse effect, Therapeutic effect, Tolerance

<table>
<thead>
<tr>
<th>Region of Body</th>
<th>Conversion Factor, EDLP (mSv mGy⁻¹cm⁻¹)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head and neck</td>
<td>0.0031</td>
</tr>
<tr>
<td>Head</td>
<td>0.0021</td>
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<tr>
<td>Neck</td>
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<tr>
<td>Chest</td>
<td>0.014</td>
</tr>
<tr>
<td>Abdomen and pelvis</td>
<td>0.015</td>
</tr>
<tr>
<td>Trunk</td>
<td>0.015</td>
</tr>
</tbody>
</table>


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\[
L_2 = 2L_1 \quad \text{and} \quad \text{CTDI}_{\text{vol1}} = \frac{\text{CTDI}_{\text{vol2}}}{2}
\]

\[\therefore \text{DLP}_1 = \text{DLP}_2\]

**FIGURE D.54**  Examples of calculating dose length product.
Dose limits
(Radiation Protection) The following dose limit values apply to exposure from practices, with the exception of exposures from medical practices (diagnostic and therapy) and from natural sources.

**Dose Limits for Occupational Exposures:** Occupational exposures of any worker shall be controlled in order not to exceed the following limits: (1) effective dose to the whole body: 20 mSv per year averaged over a period of 5 years with an effective dose of 50 mSv in any single year; (2) effective dose to the embryo or foetus: 1 mSv; (3) annual equivalent dose to the lenses of eye: 150 mSv and (4) annual equivalent dose to the skin and extremities: 500 mSv.

For occupational exposure from radionuclides, with risk of internal contamination, there are tables (for the various radioisotopes) that give ingestion and inhalation dose coefficients. This means that it is possible to evaluate the committed effective dose per unit intake via ingestion corresponding to different gut transfer factors, for various chemical forms, and the committed effective dose per unit intake (via inhalation) for the lung absorption, considering also the component cleared from the lung to the gastrointestinal tract.

In the case of apprentices of 16–18 years old, who are under training with activities involving exposures, the following limits shall be respected: an effective dose to whole body of 6 mSv per year; an annual equivalent dose to the lenses of eye of 50 mSv and an annual equivalent dose to the extremities and skin of 150 mSv.

There might be ‘special circumstances’ under which a temporary change in dose limitation is required and approved. In these cases the 20 mSv shall be averaged over a period up to 10 years; with the maximum dose per year remaining 50 mSv (circumstances shall be reviewed when any worker reaches 100 mSv). Alternatively, the annual limit shall not exceed 50 mSv with a temporary change period not exceeding 5 years.

**Dose Limits for Public Exposures:** The averaged estimated doses to critical groups of the public, attributable to practices shall not exceed the following limits: (1) effective dose to the whole body: 1 mSv per year, in special circumstances averaged over a period of 5 years with an effective dose of 5 mSv per year; (2) annual equivalent dose to the lenses of the eye of 15 mSv and (3) annual equivalent dose to the skin of 50 mSv.

**Dose Limits for Comforters and Visitors of Patients:** The dose of any voluntary comforter or visitor of patients shall be constrained so that it will not exceed 5 mSv during the period of the diagnostic investigation or the treatment. The dose to children visiting patients who have ingested radioactive material, shall be limited to less than 1 mSv.

**Further Reading:** International Basic Safety Standard for Protection against Ionizing Radiation and for the Safety of Radiation Sources, Safety Series No. 115. 1996. International Atomic Energy Agency, Vienna, Austria.

Dose model evaluation
(Radiation Protection) Dose response curves such as the linear response curve, threshold dose response curve and non-linear response curve, etc. are used either separately or in combination as models to describe the response of the human body to exposure to various types on ionising radiation from both internal and external exposure, and at high and low doses and dose rates. Such responses are dependent on the cellular responses to such different types and energies of ionising radiation.

These radiobiological models range from the very simple – e.g. the linear no-threshold model, which assumes that the response to exposure to ionising radiation is proportionate, and for which there is no threshold – to the more complex to take into account observed data on bystander effects and adaptive responses. In some circumstances the different responses may compete against each other.

It is therefore necessary to continue to observe immediate acute and late chronic effects of radiation exposure to evaluate the dose models in order to more appropriately apply them to specific exposure situations.

For more information, see the articles on Radiobiological model, Bystander effects and Adaptive responses.

**Related Articles:** Radiobiological model, Bystander effects, Hormesis

Dose modulation
(Radiotherapy) A multileaf collimator permits the modulation of the photon fluence by the motion of jaws and leaves and these results in a dose modulation in the irradiated volume.

Dose monitoring
(Radiation Protection)

**Individual Monitoring and Exposure Assessment:** Registrants and licensees, employers of workers as well as self-employed individuals shall be responsible for the occupational exposure of workers, with individual monitoring when appropriate, and shall assure that adequate arrangements be made for appropriate dosimetry and services.

Individual monitoring shall be provided, when appropriate, to workers, normally or occasionally employed in controlled areas, who may receive significant occupational exposures. In case individual monitoring is not feasible or inappropriate, the exposure of the workers should be assessed on the basis of the results of the workplace monitoring and information on duration and location of each worker.

Individual monitoring shall not be provided to workers employed in a supervised area or who enter only occasionally a controlled area. The occupational exposure of this kind of workers shall be assessed on the result of the workplace monitoring, location and time require for the assigned work.

The frequency, precision, nature and therefore choice of dosimeters shall be made considering the nature of the exposure, likelihood and magnitude of consequent potential exposure and risk.

The effectiveness of the protection devices provided, shall be demonstrated, with particular attention to the respiratory equipment used by workers who may be exposed to radioactive contamination. The intake of radioactive substances or the committed dose, as appropriate must be assessed.

**Monitoring of the Workplace:** Registrants and licensees shall establish, maintain and keep under review a programme for monitoring of the workplace, under the supervision, (following the requirements of the regulatory authority) of qualified experts or radiation protection officers.

The nature and frequency of monitoring shall

1. Enable (a) to evaluate the radiological conditions in all workplaces, (b) to assess the exposure in controlled and supervised areas and (c) to review the classification
2. Depend on the level of ambient dose equivalent and activity concentration, taking into account fluctuations, likelihood and magnitude of potential exposures
The programme shall specify

1. Quantities to be measured
2. Frequency and position of measurements
3. Appropriate methods and procedures
4. Reference levels and actions to be taken if they are exceeded

If appropriate, records of the finding shall be kept and made available also to the workers.

Health Surveillance: In accordance with the rules established by the National Authority, the employers, registrants and licensees shall make arrangements for appropriate health surveillance. Health surveillance shall be based on general principles of occupational health and designed to assess the initial and continuing fitness of the workers for their respective assignments.

Records: Employers, registrants and licensees shall maintain exposure records for each worker described earlier under individual monitoring and exposure assessment.

The exposure records shall include

1. Information on the nature of work
2. Doses, exposures, intake values and data upon which the dose assessment has been based
3. Information on dates of employment and relative doses, exposures and intakes in each employment, in case of a worker who is or has been occupationally exposed with more than one employer
4. Records of any doses, exposures or intakes due to emergency interventions or accidents, which shall be distinguished from those during normal work

Employers, registrants and licensees shall

1. Provide for access by the workers to their own records
2. Provide for access by the supervisor of the health surveillance programme by the regulatory authority and relevant employer
3. Facilitate the provision of copies to new employment when necessary
4. Make arrangements for the retention of record, as appropriate, when a worker cease to work
5. Ensure maintenance and confidentiality of data

Exposure records for each worker shall be kept during the worker’s working life and afterwards until the worker attains or would have attained the age of 75 years, anyhow for not less than 30 years.


Dose optimisation (Radiotherapy) Dose optimisation in radiotherapy is a treatment planning procedure in which the treatment beam parameters and machine parameters are adjusted so as to optimise the dose and dose distribution to better meet with the dose prescription. In planning of conventional treatment, dose optimisation is a manual process, in which the planner manually modifies the beam parameters such as beam angle, wedge angle, beam size, and beam energy to achieve the dose specification. In IMRT planning, dose optimisation is achieved through inverse planning (see Inverse treatment planning).

Dose profile (Diagnostic Radiology) In CT the term ‘dose profile’ is used to describe the dose distribution, along the axis of rotation (z-axis), from a single sequential irradiation. The irradiated ‘slice’ is defined as the full width at half maximum (FWHM) of this distribution.

The dose profile should not be confused with the slice, or z-sensitivity profile. The width and shape of the dose profile are determined by the primary collimator opening and geometry, and the focal spot size (Figure D.55).

Figure D.55 shows the dose profiles resulting from different configurations in green. (a) an ideal rectangular profile from a point source, (b) a realistic trapezoidal profile caused by the penumbra produced by a finite focus, (c) is for a similar configuration as (b) but with a reduced collimator opening and (d) the increased penumbra resulting from collimators closer to the focus.

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**FIGURE D.55** Dose profiles along axis of rotation (z-axis)—(a) an ideal rectangular profile from a point source; (b) a realistic trapezoidal profile caused by the penumbra produced by a finite focus; (c) profile for a similar configuration as (b) but with a reduced collimator opening; (d) profile influenced by the increased penumbra resulting from collimators closer to the focus.
Dose profiles are generally measured in air, on the scanner’s axis of rotation, with x-ray film or radiochromic film. They can also be measured in a phantom with thermoluminescent dosimeters (TLDs). Figure D.56 shows once such profile obtained with radiotherapy verification x-ray film and scanned on a scanning microdensitometer. The optical densities have been converted to exposure readings by using a calibrated film.

**Related Articles:** Slice thickness, Slice profile

### Dose profile

(Substitute) The profile of a photon beam consists of two distinct regions, the umbral and the penumbral region. The umbral region is where the beam profile is unaffected by the collimators while in the penumbral region the field defining collimators affect the beam profile. There is no exact indication where the transition between the two regions occurs but in general a nominal position from 1 to 1.5 cm inside the geometric field edge, which is considered the 50% dose level, is generally accepted as the approximate location where the transition occurs. In the penumbral region the primary photons are shielded by the collimator jaws and therefore in this region there is a sharp gradient fall-off, sigmoid in shape also because of the small contribution of the extra focal radiation. The width of the penumbral region is some millimetres due to the finite size of the virtual x-ray source and lateral electron disequilibrium. The penumbral region has a long shallow fall off in dose due mainly to the scatter radiation and less due to transmission through the jaws. The scatter radiation increase with depth and field size and subsequently the penumbral dose in the tail increase in magnitude with depth and field size.

**Related Articles:** Penumbra, Radiation field, Beam flatness

### Dose rate

**Radiation** Dose rate is defined as the absorbed dose per unit time. The dose rate can be used to estimate a biological effect provided the effect is proportional to the dose rate. If an equal radiation dose is received over different time intervals, different biological effects are observed. Fractionated irradiations allow damaged cells to repair between irradiations (classic dose rate effect).


### Dose rate constant

**Nuclear Medicine** The dose rate constant, $\Gamma$ is an estimation of the effective dose received from x-rays and $\gamma$-rays from external sources. $\Gamma$ is measured at 1 m distance and the unit is mSv m²/MBq h. $\Gamma$ depends on nuclide characteristics like the number of x-rays and $\gamma$-rays per disintegration, their energies and the attenuation properties of the tissue. Generally low energy (typically $<10$ keV) x-rays and $\gamma$-rays are excluded from the dose rate constant calculations since they do not pose any greater hazard because of their low penetration through syringe and vial walls.


### Dose rate dependence

**Radiation** The response of tumours and normal tissues to radiation depends on the rate of dose delivery: a reduction in the dose rate is generally associated with a reduction in the radiation response (although a few exceptions to this rule have been noted).

In external beam radiotherapy, dose is typically delivered at dose rates of 1–5 Gy/min and the linear quadratic model has been shown to describe the response to radiation both in vitro and in vivo well. However, as dose rate is lowered, the time taken to deliver the radiation dose is extended and it becomes possible for the radiation response to be modified as a result of the following processes: repair of sub-lethal damage, redistribution, repopulation and reoxygenation. Such considerations are required in the clinical situation of LDR brachytherapy and in targeted radiotherapy.

Repair is the fastest of the four processes. When exposure is on the order of its half-time (~1 h), considerable repair will take place. Therefore, repair will modify radiation effects over the dose-rate range 0.001–1 Gy/min. Repopulation is a much slower process with doubling times for human tumours and normal tissues ranging from a few days to a few weeks. Hence, significant repopulation will only occur during a radiation exposure when its duration is a significant fraction of a day which requires a dose rate below 2 cGy/min, depending on the cell proliferation rate. Redistribution and reoxygenation will modify response over an intermediate range of dose rates.

As the dose rate is lowered in the range from 1 Gy/min down to 1 cGy/min, the radiosensitivity of cells decreases and the shouldered cell survival curves observed at high dose rate gradually become straighter as only lethal damage is expressed due to the repair of sub-lethal damage. This is illustrated in Figure D.57 where curve A, typical of that obtained at a dose rate of around 1 Gy/min, has a marked curvature compared with curve B, typical of that obtained

at a dose rate of around 1 cGy/min. The dose recovery factor (DRF) provides a measure of the amount of sparing associated with the reduction in dose rate and is the ratio of the doses that give a specified surviving fraction: $D_k/D_o$ in Figure D.55. The dose needed for 1% survival is around one to three times higher at 1 cGy/min than at 1 Gy/min.

Most normal tissues show considerable sparing as the dose rate is reduced. For example, a study by Down et al. (1986) on mouse lung irradiated with Cobalt-60 γ-rays found that the DRF for acute pneumonitis, measured at an incidence level of 50%, was 2.6 when the dose rate was reduced from 100 to 2 cGy/min. It was not possible in this experiment to reduce the dose rate below 2 cGy/min but when the incomplete repair model (see article on Linear quadratic (LQ) model) of Thames (1985) is fitted to the data, it predicts a DRF of 4.4 when the dose rate is reduced further to 0.01 cGy/min (again compared with the dose rate at 100 cGy/min). It is expected that the proliferation of stem cells in the lung will produce even greater sparing at very low dose rates.

In some situations an ‘inverse dose-rate effect’ has been observed which has been attributed to the low dose rate allowing the progression of cells through the cell cycle into the more sensitive phases and thereby suffering greater damage. A further lowering of the dose rate will allow cells to escape the radiosensitive phases and this combined with the effect of cell proliferation will produce a shallower survival curve.

**Abbreviation:** DRF = Dose recovery factor.

**Related Articles:** Cell cycle, Cell proliferation, Cell survival curves, Dose response model, Linear quadratic (LQ) model, Radiosensitivity, Redistribution, Reoxygenation, Repair, Repopulation, Surviving fraction


**Dose rate distribution**

*(Radiotherapy)* The dose rate distribution is particularly important in brachytherapy. It is a map of the dose rate around a radioactive source. The dose rate distribution depends on radionuclide decay properties and the activity of the source. It can be used to calculate the absorbed dose in different locations around one or several point sources.

**Dose rate effectiveness factor (DREF)**

*(Radiation Protection)* The dose rate effectiveness factor (DREF) is used to estimate the stochastic biological effects from exposures to ionising radiation that are spread over time. This allows us to account for effects of healing mechanisms within the body that reduce the implications of a radiation dose when applied at a low dose rate. For example, an instantaneous dose of 1 Sv to a human produces an approximate risk of 5% of a fatal cancer developing. But, if that dose was spread across a longer time span (from days to weeks or years), then the risk would be reduced by a factor dependent on the timescale of exposure.

In general a factor of between 2 and 10 would be expected, depending on the timescale. However, for radiation protection purposes, and to err on the side of caution, a factor of 2 is used for all circumstances.

**Hyperlinks:** EMERALD (DR module), www.emerald2.eu

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**Dose reference point**

*(Radiotherapy)* This is generally the point at which the prescription dose is delivered. In brachytherapy, e.g. a point 0.5 cm directly out from the surface of the ovoid pair along the left to right axis of the pair is often used as the dose reference point (Figure D.58). Another example also in brachytherapy is Point ‘A’ in the Manchester system. In external beam radiotherapy, the isocentre is most often the dose reference point for prescription (ICRU report 29).

Dose reference points can also be used to compare doses to critical organs between different prescribing systems. An example of this is the International Commission on Radiation Units and Measurements (ICRU) bladder point in intra-cavitary brachytherapy (ICRU report 38). This point is determined by taking radiographs with a radio-opaque balloon in the bladder prior to treatment. A point at the centre of the balloon is marked and the dose recorded.


**Dose response curve**

*( Radiation Protection)* Dose response curves such as the linear response curve, threshold dose response curve and non-linear response curve, etc. are used either separately or in combination as models to describe the response of the human body to exposure to various types of ionising radiation from both internal and external exposure, and at high and low doses and dose rates.

This response may be described mathematically as the dose response function.

For more information, see the articles on Linear no-threshold dose model, Stochastic effects and Non-stochastic effects.

**Related Articles:** Linear no-threshold dose model, Stochastic effects, Non-stochastic effects

**Dose response function**

*(Radiation Protection)* The response of the human body to differing doses of ionising radiation may be described mathematically as the dose response function, which can then be plotted graphically as a dose response curve.

For more information, see the articles on Dose response curves, Stochastic effects and Non-stochastic effects.

**Related Articles:** Dose response curve, Stochastic effects, Non-stochastic effects

**Dose response model**

*(Radiotherapy)* Dose response models are mathematical models that describe radiobiological phenomena pertaining to the response of
mammalian cells (tumour or normal tissue) to radiation. In general, dose response models describe how the probability or frequency of a specific response changes with the dose. Such models are usually derived from the fitting of mathematical functions to data obtained either from laboratory experiments on cell cultures, animal experiments, or human data obtained either by studying patients who have received radiation therapy or those involved in radiation accident/ incidents such as the victims of Hiroshima and Chernobyl. Examples of such models include the linear quadratic model, tumour control probability and normal tissue complication probability.

Related Articles: Linear quadratic (LQ) model, Normal tissue complication probability, Radiobiological models, Tumour control probability

Dose tolerance

(Radiotherapy) Radiation treatment inevitably affects normal tissue and so may cause radiation induced adverse effects. In radiotherapy, it is generally the case that the total dose that can be tolerated depends on the volume of tissue irradiated – the dose-volume effect. Additionally, the tissue architecture is thought to be important in determining the tolerance dose for partial organ irradiation. For further information, see the articles on Tolerance.

Related Article: Tolerance

Dose verification

(Radiotherapy) Verification is a key component of radiotherapy. The purpose is to ensure that the treatment is geometrically and dosimetrically accurate, delivering the desired high dose to the target and achieving the required dose limits to normal tissues. Dose verification may be done prior to treatment or during treatment.

Pre-treatment Dosimetric Verification: The planned treatment beam arrangement is transferred to a model of a phantom in the TPS, and a comparison between the prediction of the TPS and the dose measured in the phantom is made. Point dose and 2D dose (or 3D dose) distribution should be verified. For the verification of the point dose ionisation chambers (Farmer type or PinPoint chamber) can be used (Figure D.59). For relative dose distribution films, radiochromic films, 2D dosimeter arrays (2D matrix of detectors), EPID, or radiosensitive gel can be used.

Dosimetric Verification during Treatment (In Vivo Dosimetry): This involves measuring the dose at points where the beam enters the patient (usually diodes or TLD), where it exits the patient, or in an anatomical cavity. Other approaches involve measuring the dose in a plane outside the patient using a detector such as an EPID. This is called in vivo dosimetry.

Abbreviations: TPS = Treatment planning system, TLD = Thermoluminescent dosimeter, EPID = Electronic portal imaging device and IMRT = Intensity modulated radiation therapy.

Related Articles: Field verification, Treatment verification, Detectors, Treatment planning system, Electronic portal imaging device, Intensity modulated radiation therapy

Dose volume histogram

(Radiotherapy) A dose volume histogram (DVH) is a tool used in radiotherapy to aid evaluation of treatment plans. It provides a summary of the dose distribution throughout a specifically defined volume of interest, e.g. the target volume or an organ at risk (OAR). There are two alternative forms of DVHs, as shown in Figure D.60. A differential DVH shows the total volume of voxels receiving a dose in a specified dose interval, against a set of equally spaced dose intervals. Alternatively, a cumulative DVH shows the total volume of voxels receiving a dose greater than, or equal to, a specified dose, against dose. This is the format most widely used in radiotherapy. The volume and dose can either be specified as absolute units, or as percentages (either of maximum or prescribed dose).

DVHs are helpful in assessing the dose uniformity across the target volume, as well as evaluating if the dose received by an OAR is above the tolerance value. This is especially useful for evaluating the dose received by a parallel organ, such as the lung or liver, where dose limits to normal tissue are often expressed in terms of the percentage volume receiving a specified dose. They can be used as a graphical way of comparing multiple treatment plans on a single graph. An ‘ideal’ cumulative DVH for a target volume and a critical volume are shown in Figure D.61. However they must be considered alongside the spatial dose distribution for a complete picture of the treatment plan.

They are also used as input data to obtain statistical measures such as tumour control probability (TCP) and normal tissue

FIGURE D.59 Farmer ionisation chamber. (Graphs courtesy of EMERALD project, www.emeral2.eu)

FIGURE D.60 Indication of how DVHs are constructed.

FIGURE D.61 Examples of integral DVHs for PTV (a), and OAR (b). The solid line shows the ideal situation, whereas the dotted line is more realistic.
Dose volume histogram (DVH), differential (Radiotherapy) See Dose volume histogram

Dose volume histogram (DVH), integral cumulative (Radiotherapy) See Dose volume histogram

Dose width product (DWP) (Diagnostic Radiology) The dose width product is a measure of the radiation output, usually expressed as absorbed dose in air (or air kerma), multiplied by the width of the radiation beam. This measure is particularly useful for narrow slit beams of radiation and is considered the most appropriate quantity for the assessment of patient dose with panoramic or orthopantomograph (OPG) dental units. Diagnostic reference doses for OPG units are expressed in terms of the DWP measured in mGy.nn.

The DWP represents a measure of the total beam energy across the beam width independent of the exact shape of the beam profile. The DWP of a square profile can simply be calculated from the absorbed dose measured within the beam, using TLD or a small area solid state detector, multiplied by the full width half maximum of the beam profile usually determined with the use of film. If the beam has a more Gaussian profile then the DWP can best be calculated from the total dose across the beam width as measured with an isotropically sensitive detector, such as a CT pencil chamber, multiplied by the chamber length.

Dosemeter (Radiotherapy) The specific use of the terms dosimeter and dosemeter in English refers to the following common meanings:

Dosimeter is a passive device used for patient or staff dosimetry. This device accumulates the dose over a certain period of time and is retrospectively read. Example of dosimeters are film badges, thermoluminescent dosimeters, etc. Each type of dosimeter requires specific calibration.

Dosemeter is an active device which measures dose (or dose rate) in real time. Example of dosemeter is ionisation chamber with electrometer.

Related Articles: Dosimetry, Film badge, Finger ring dosimeter, Thermoluminescent dosimeter (TLD)


Dosimetry (Radiation Protection) Dosimetry, in its meaning of measurements of doses and related quantities is at the basis of radiation protection as it is essential ‘to measure in order to protect’. Primarily it is essential to measure in order to use the appropriate quantity/quality of radiation for diagnostic or therapy applications. Dosimetry covers a vast field including all aspects of personal and environmental dosimetry and monitoring for both ionising and non-ionising radiation; involving biological, physical and biophysical aspects. It includes also external and internal personal dosimetry for patients and staff, environmental and personal monitoring (see also Occupational exposures), workplace monitoring and accidental dosimetry. The definition of adequate measurement units is therefore essential in relation to the different kind of radiations. An appropriate dosimetry programme includes methodology of measurements and appropriate units based on the know-how of the kind of radiation beam, which is going to be measured and its interactions. In this article, only as an example, the most usual radiation quantities and units needed in the field of diagnostic radiology and corresponding dosimetry are mentioned.

Equivalent dose is helpful in quantifying the potentially harmful biological effects of the absorbed dose, modified by the radiation weighting factor, the SI unit is the sievert (formerly rem); it relates to the biological effect of radiation in an organ or tissue. The concept of effective dose allows expressing the detriment resulting from stochastic effects following exposure of different tissues or organs. Effective dose is a weighted sum of equivalent organ doses according to tissue specific weighting factors ($W_T$ values are given), and it is useful to estimate the stochastic risk.

The purpose of patient dosimetry is to assess doses (at the entrance or skin or in depth), usually with backscatter (present in the actual examination). Several concepts are used according to measuring technique: the incident dose (ICRP definition), entrance skin dose, backscatter factor (BSF), DAP (dose-area product available on special devices).

In the field of mammography dose is expressed in terms of average glandular dose.

In CT specific concepts are used to assess dose from an examination: multiple scan average dose, CT dose index, dose-length product. Normalised indices can be defined to allow comparison with reference values to be made.


Dosimetry protocol
(Radiotherapy) The dosimetry protocol (also known as a code of practice) defines the procedures to follow for the dosimetric calibration of clinical beams either in radiotherapy or in diagnostic radiology. These protocols are written at the national (e.g. IPEM [United Kingdom], AAPM [United States], DIN [Germany], etc.) and international level (e.g. IAEA TRS 398, IAEA TRS 457). These protocols are written also in association with primary standard laboratories which ensure high level of consistency throughout departments and among different countries.

Abbreviations: AAPM = American Association of Physicists in Medicine, DIN = Deutsches Institut für Normung, IAEA = International Atomic Energy Agency and IPEM = Institute of Physics and Engineering in Medicine.

Related Articles: Calculation of absorbed dose, Electron beam dosimetry, Gray (Gy), Beam quality, Quality index, Reference depth, Dosimetry report

Dosimetry report
(Radiotherapy) A dosimetry report can mean a number of different things.

It can be used to describe the results from an independent dosimetry audit of a radiotherapy department’s equipment. This will relate the dosimetry values quoted by the department to those measured by the external assessor using a different set of measuring instruments.

Some people may use it to describe the documentation received from a primary or secondary dosimetry standard laboratory, although this is more commonly known as a calibration certificate.

A dosimetry report can also be used when an error in treatment has been discovered to provide an assessment of the dosimetric impact, as a part of the protocol on accident.

A further use of the term dosimetry report is in the area of personal dosimetry, where the results from the person’s film badge or TLD are documented over a set period of time (e.g. results for each month of a year).

Dosimetry systems
(Radiotherapy, Brachytherapy)

Dosimetry Systems for Brachytherapy: Before the time of computers, practical means of performing brachytherapy relied on agreed rules based on clinical experience, pre-calculated dose distributions, and a predictive dosimetry system. The coming of image guided brachytherapy has enabled true definition of volumes of interest, as well as three-dimensional dose distribution calculations causing a decline in the use of classical methods of dose prescribing. However it should be noted that there still exists a wealth of clinical experience that has been gained using these historical ‘systems’. Although image guided brachytherapy is a huge step forward, care must therefore be exercised when changing from ‘classical’ to new image-based dose prescribing systems.

Systems for Treatment of Cervix Carcinoma: In ICRU Report 38 ‘Dose and Volume Specification for Reporting Intracavitary Therapy in Gynaecology’ the following terminology is used: ‘the term “system” denotes a set of rules taking into account the source strengths, geometry and method of application in order to obtain suitable dose distributions over the volume(s) to be treated. For reporting, the system includes recommendations for specifying the application and possibly, as in the Manchester system, for calculating the dose rate (or dose) at specific points.’

Three classical radium-based systems, all based on clinical experience, are presented in short in this report:

- The Stockholm system, ‘intrauterine probe plus plate’
- The Paris system, ‘intrauterine probe plus ovoids’
- The Manchester system, ‘intrauterine probe plus ovoids’

Examples of applications for these three systems are also given.

Today (2009) radium is no longer used as a brachytherapy source, and the manual radium loading techniques have been replaced with remote controlled afterloading systems using 137Cs (low dose rate) and 192Ir sources (high dose rate and pulsed dose rate).

Many of the applicators used today for cervix cancer brachytherapy are based on the classical systems, e.g. the ring applicator which is developed from the Stockholm system, see the article on Intracavitary brachytherapy.

Systems for Interstitial Brachytherapy: ICRU Report 58 ‘Dose and Volume Specification for Reporting Interstitial Therapy’ states: ‘The term “system” denotes a set of rules which takes into account the source types and strengths, geometry and method of application to obtain suitable dose distributions over the volume(s) to be treated. The system also provides a means of calculating and specifying dose. It is important to remember that while an implant may follow the source distribution rules of a system, it does not comply with the system unless the method of dose specification and prescription are also followed. In addition, if the implant rules are modified, the dose uniformity intended by the system may be compromised.’

Three classical interstitial systems are mentioned:

- The Manchester, Paterson–Parker, system (Ra)
- The Quimby system (Ra)
- The Paris system (192Ir wires, LDR)

Today (2009) image guided techniques are used more and more for interstitial brachytherapy, e.g. the ultrasound guided interactive permanent seed implantations for prostate cancer brachytherapy (see the articles on Interactive implant technique and interstitial brachytherapy).

For an overview of dosimetry systems, both for intracavitary and interstitial brachytherapy, the reader is referred to the GEC ESTRO Handbook of Brachytherapy, and references therein.
Double contrast

(Diagnostic Radiology) A double contrast study of the Gastro intestinal tract uses a small amount of barium meal to coat the surfaces and then air or gas to distend the colon for better visibility of potential lesions.

Double exposure radiograph

(Diagnostic Radiology) A radiograph that has been exposed a second time by error.

Double focus tube

(Diagnostic Radiology) Usually an x-ray tube producing power exposures would require large filament coil wire. However such tube could not produce small effective focal spot, necessary for high resolution radiographs. In this case two wires are used (large and small) allowing two effective focal spots. Usually rotational anode x-ray tubes use double focus.

There are two main constructions of double focus x-ray tubes – with the filaments one to the other and with the filaments one above the other. In the first case both foci overlap, but in this case the actual focal spots also overlap, causing overheating of some part of the anode surface and leading to shorter lifetime of the x-ray tube (Figure D.62). In the second case both filaments have different actual focal spots, but the central beam is displaced which should be taken into consideration in centring the radiographs.

Note on Figure D.62 that projections of both cathode filament coil wires (for large/broad focus BF and for small/fine focus FF) overlap over part of the actual focal spot, which appears overloaded.

Related Articles: Stationary anode, Rotating anode, Target, Line focus principle, Biangular anode disk, Focal spot actual, Focal spot effective, Focal spot

Downscatter

(Nuclear Medicine) Downscatter refers to an effect that is common when acquiring simultaneous transmission and emission images. The transmission image is used for attenuation correction and emission images are acquired using a different radionuclide, hence a different emission energy. The two scans can therefore be acquired by counting events in two different energy windows. Inevitably a number of photons from the high energy emitting radionuclide undergo Compton scattering in the object before being registered by the detector. The energy registered in such an event is lower than the original emission energy and might be falsely registered in the lower energy window. Another cause of downscatter is poor crystal energy resolution and/or small difference in emission and transmission photon energy, such that events in the two windows will be ‘misplaced’ and they are registered in the ‘wrong’ window.

Downscatter can also be a problem when acquiring the scans sequentially because the radiopharmaceutical is sometimes administered prior to the examination (sometimes several hours in advance) and activity is therefore present during the transmission scan as well. Downscatter can be estimated and corrected for in the same fashion as scatter correction (see separate article). An energy window between the transmission and emission windows can be used to estimate the level of downscatter.

Related Articles: Positron emission tomography (PET), Scatter correction in SPECT, Single photon emission computed tomography (SPECT)


DQE

(Diagnostic Radiology) See Detective quantum efficiency (DQE)

Drift

(Radiation Protection) The movement of charge carriers produced by ionising radiation (electrons and holes in semiconductor devices, ions and electrons in gas-filled radiation detectors e.g. ionisation chamber) under influence of the external electric field \( E \) [V/m] created between the electrodes is called drift. The drift current density \( J \) [A/m²] is equal to the sum of the negative charge carriers (electrons, negative ions) \( J_n \) and positive carriers (holes, positive ions) \( J_p \):

\[
J = J_n + J_p
\]

where

\( J_n = n_n q_n v_n \) and \( J_p = n_p q_p v_p \),

\( n_n, n_p \) are the number of negative and positive carriers per cm³, respectively

\( q_n \) and \( q_p \) are the electric charge of the carriers

\( v_n \) and \( v_p \) are their respective velocities
Introducing the mobility \( u [m^2/(V \cdot s)] \) of a charge carrier as a ratio of its velocity \( V \) and electric field strength \( E \), the dependence between current density and electric field \( E \) can be expressed as

\[
J = (n \cdot q \cdot u_n + p \cdot q \cdot u_p) \cdot E
\]

The net velocity of the charge carriers is a combination of the thermal velocity (i.e., about \( 10^3 \text{cm/s} \) for silicon at room temperature) and the drift velocity produced by the applied electric field \( E \). The mobility of charge carriers remains constant and small compared to the thermal velocity over wide ranges of electric field strengths and is solely dependent on the kind of detector.

**Related Articles:** Gas-filled radiation detectors, Semiconductor detector


**Drift mobility**

(Radiation Protection) The external electric field applied to the radiation detector (e.g., gas-filled chamber or semiconductor) produces a drift. The free charge carriers (electrons, positive and negative ions in gas-filled radiation detectors, electrons and holes in semiconductor detectors) are accelerated in the direction determined by the electric field.

The drift mobility \( u \) for ions in gas-filled detectors is equal to

\[
u = \frac{V \cdot p}{E} \quad [m^2 \cdot \text{atm}/(V \cdot s)]
\]

where

\( V \) [m/s] is the drift velocity of ions

\( p \) [atm] is the gas pressure

\( E \) [V/m] is the electric field strength

The values for mobility of positive and negative ions are between 1 and \( 1.5 \times 10^{-4} m^2 \cdot \text{atm}/(V \cdot s) \) in the gas at 1 atm pressure and with an applied electric field \( E \) about \( 10^5 \text{V/m} \). The mobility of free electrons in the same gas is about \( 10^5 \) greater because of their much lower mass.

The drift mobility for electrons \( u_n \) and holes \( u_p \) is defined as

\[
u_n = \frac{v_n}{E} \quad [m^2/(V \cdot s)]
\]

\[
u_p = \frac{v_p}{E} \quad [m^2/(V \cdot s)]
\]

where

\( v_n \) and \( v_p \) [m/s] is the velocity of electrons and holes, respectively

\( E \) [V/m] is the electric field strength

The following table demonstrates how the drift mobility of electrons and holes in silicon and germanium is affected by the temperature of the semiconductor:

<table>
<thead>
<tr>
<th>Temperature (K)</th>
<th>Silicon (cm²/(V s))</th>
<th>Germanium (cm²/(V s))</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Electron</td>
<td>Hole</td>
</tr>
<tr>
<td>300</td>
<td>1,350</td>
<td>480</td>
</tr>
<tr>
<td>77</td>
<td>21,000</td>
<td>11,000</td>
</tr>
</tbody>
</table>

As shown in the table, the drift mobility of electrons and holes in both silicon and germanium is significantly increased by reducing the temperature of the material; from ambient room temperature (~300 K) down to that of liquid nitrogen (77 K), improving the performance of such materials when used as radiation or imaging detectors.

**Related Articles:** Fast imaging with steady state (FISP), PSIF (FISP reversed), SSFP, Steady state free precession, \( T_2 \)-weighted


**Dual energy digital radiography**

(Diagnostic Radiology) Dual energy digital radiography is a new technique used to improve the visibility of details primarily in chest imaging although it can also be employed in other examinations.
Technically, two images are acquired at different x-ray energies. Soft tissue and bone have different absorption properties depending on the beam spectrum. This aspect is utilised to produce three separate images: a conventional planar image and, derived from this, an image with only soft tissue and an image with only bone and calcified tissue.

Two different technical solutions exist to discriminate soft tissue and bone, depending on the digital detector used.

- Using photostimulable phosphor plates, a single exposure is taken using a dual detector made by two integrated CR plates, separated by a thin copper filter. This technique avoids the risk of blurring due to patient movement, but the back plate receives a lower dose than the front plate and this increases the noise in this image.

- Using a direct digital detector two exposures are taken in rapid succession, selecting different kV values. This solution is possible only on equipment able to acquire two images in a very short temporal interval. It offers the advantage to select very different x-ray energies (typically the consecutive kV selections are 60 and 120 kV) and to produce images with a lower noise level. The disadvantage is the risk of voluntary and especially involuntary patient movement between the exposures.

**Dual energy imaging**

(Diagnostic Radiology) In CT scanning dual energy (dual-energy) imaging refers to the acquisition of images at two different x-ray energies. This enables improved differentiation and characterisation of materials in the imaged object. Conventional CT scanning is performed with a single energy spectrum of x-rays. The information obtained is of the distribution of linear attenuation coefficients and does not uniquely identify the materials.

Dual-energy CT is usually achieved by measuring the attenuation using two different energy spectra, e.g., one at 80 kVp and the other at 140 kVp, but can also be performed with one energy spectrum and a dual-layered detector. For dual x-ray spectra acquisition, one approach is the use of a dual source system, with the two x-ray tubes operating at different kilovoltage potentials (kVp). Another method, with a single source system, is to switch the tube kVp between alternate projections within a rotation. The two data sets can then be analysed for differences in attenuation enabling a better characterisation of selected materials.

One clinical application of dual-energy CT is better differentiation between bone and iodine contrast material, allowing for improved bone removal in CT angiography images. Some other applications include diagnosis of kidney stone type, and discrimination between calcified plaque and contrast material in coronary angiography.

**Related Articles:** Computed tomography, Dual source CT

**Dual energy subtraction**

(Diagnostic Radiology) Dual energy subtraction is an x-ray imaging method that can be used to separate tissues, especially soft tissues and bone. Because of its calcium content, bone has a relatively high x-ray attenuation at the lower photon energies. The difference between bone and soft tissue attenuation is less at the higher photon energies.

A typical procedure would produce 2 images with 2 kV values, such as 60 and 120 kV.

An image subtraction can then be used to produce both bone-selective and soft tissue-selective images in addition to conventional radiographs (Figure D.64).

**Dual filament tube**

(Diagnostic Radiology) Each classical x-ray tube with two effective focal spots requires two filament wires – a large filament wire (producing broad focus) and a small filament wire (producing fine focus). Usually these dual filament tubes are with rotational anode (including all biangular anode tubes). See Figure D.62 and Figures R.56 through R.59.

These two wires can sit in one focussing cup, or in different focussing cups (Wehnelt electrode) – see Figure C.16. The two wires can be positioned one to the other or one above the other. In the first case the two filaments create overlapping of actual focal spots. This could lead to cracks (due to thermal stress), and will decrease the life of the tube. When the two filaments are one above the other, this thermal problem is avoided, but the two effective focal spots are slightly displaced, which should be corrected at beam centring.

**Related Articles:** Cathode, Filament, Rotating anode, Double focus tube, Target, Biangular anode disk, Focal spot actual, Focal spot effective

**FIGURE D.64** A conventional radiograph (left) compared to a bone-selective image produced with a dual energy subtraction method.
Dual source CT

Dual source CT became available in 2006, as the Siemens Somatom definition. The dual source scanner is based on third-generation technology but incorporates two x-ray tubes and detector banks mounted at 90° to each other (Figure D.65).

The features of dual source systems can be utilised in a number of specialist applications. The first is where a high temporal resolution is required, such as in cardiac CT scanning. Dual source systems can acquire a cardiac image in half the time of a single source system rotating at the same speed, thereby halving the temporal resolution and ‘freezing’ the motion of the heart more effectively.

Another area where dual source systems can be used to advantage is in dual energy imaging. Acquiring a scan at two different x-ray energies provides additional information on substance composition and can lead to improved tissue characterisation. An example of this is in differentiation of bone and iodine contrast. On a dual source scanner the two x-ray tubes can be operated at different kilovoltage potentials, leading to an almost simultaneous acquisition of the dual energy scans. This removes the problems of image registration encountered when the image acquisition is non-concurrent.

A third application where dual source CT can provide advantages is in situations requiring a high photon flux, such as the scanning of bariatric (obese) patients. The two tubes can provide double the power of a single source system.

A drawback of current dual source systems is that the scan plane FOV of Tube B is reduced in comparison to Tube A. However, on the latest dual source model, the Siemens definition flash (launched in 2008), it has been increased over the original dual source system.

Related Articles: Computed tomography, Dual energy imaging

Abbreviations: HDR = High dose rate and PDR = Pulsed dose rate.

Related Article: Remote after loading

Duplex ultrasound

Duplex ultrasound refers to an ultrasound device which combines a B-mode image with pulsed wave (or sometimes continuous wave) Doppler ultrasound. The position of Doppler beam can be displayed on the B-mode image, enabling positioning of the beam and, for pulsed Doppler, the sample volume to investigate a particular area of tissue. If the direction of the vessel under investigation is clear, duplex ultrasound can enable the operator to determine the beam/vessel angle, thereby permitting measurements of velocity from the Doppler spectrum.

In a duplex scanner the B-mode image may be frozen while Doppler investigation is undertaken, or the image may be updated at a user-selected interval. The B-mode image may run concurrently with the Doppler but the B-mode frame rate and the temporal resolution of the Doppler sonogram may suffer due to the limited time for each.

The term triplex scanning refers to concurrent B-mode, colour flow and spectral Doppler in a scanner. Triplex scanning has the advantage that movement of vessel can be tracked. However, the performance of each mode may be compromised (e.g. by limiting colour flow update and size of colour flow ‘box’), due to the limited amount of time available for each.

Duplex scanning first came into use in the mid to late 1970s. The term duplex is still used colloquially even in colour flow imaging and describes a scan in which B-mode and Doppler imaging is used (Figure D.66).

Duplicating film

Duplicating of x-ray radiographic film is usually made with a special duplicator – a light source where the original x-ray film is placed together with a duplicating film (usually with single emulsion). The light (often ultraviolet) from the duplicator passes through the original film (what modulates it) and after this exposes the duplicate film. This way an exact (and negative) copy of the original is produced.
Duty cycle

(Magnetic Resonance) The duty cycle is basically a measure of how heavily the system can be loaded with respect to power (heating) over time.

In a general way duty cycle can be defined as the proportion of a specified time period \((T_{on} + T_{off})\) during which a device or system is operated \((T_{on})\):

\[
\text{Duty cycle} = 100 \cdot \frac{T_{on}}{T_{on} + T_{off}} \%
\]

The way of defining the duty cycle may differ between vendors but typically it is reported to be 100% for an MRI system. A more specific description of the duty cycle than the definition used here would have to reflect its dependence on how the system dissipates power over time in relation to how the system is operated. In practise it might therefore simply be a good idea to verify the reported duty cycle by finding out if the MRI system can execute any pulse sequence with any imaging parameters for as long a time as desired without interruptions for heat evacuation.

For MRI equipment the radiofrequency (RF) and gradient systems are the ones that essentially dissipate heat and therefore needs to be cooled. A high tolerance to heating of the system is of course favourable. Water cooled gradient systems can, e.g. provide a higher duty cycle than air cooled systems, which is normally required with modern MRI scanners that often utilise fast imaging techniques.

Duty cycle

(Ultrasound) The duty cycle describes the fraction of time for which a transducer, operating in pulsed mode, transmits ultrasound. It is the ratio of the pulse length to the time between pulses. Duty cycle is also referred to as Duty factor.

EXAMPLE:

A physiotherapy ultrasound machine transmits pulses of 2 ms duration followed by 8 ms pauses. The duty cycle is 1:4. This may also be described as a duty cycle of 20%.

In diagnostic ultrasound applications the duty cycle is typically less than 1%.

Related Article: Pulsed mode

Dwell time

(Radiotherapy, Brachytherapy) A typical remote afterloading unit for high dose rate and pulsed dose rate brachytherapy has one single small source of high specific source strength with stepping movement and several treatment channels. The source movements are computer controlled, both the source step size between stop positions and the dwell time at each stop position.

Figure D.67 shows two dose distributions calculated with seven stop positions, step size 1.0 cm. In Figure D.67a, equal dwell times have been used to give a dose of 10 Gy at a distance of 2 cm from the central stop position. In Figure D.67b, the dwell times have been optimised to give 10 Gy at a distance of 2 cm from all stop positions. Notice the tapering shape in Figure D.67a and the prolonged dwell times at the first and last stop positions in Figure D.67b.

Related Article: Remote afterloading unit

Dynamic aperture

(Ultrasound) In a linear array system, the receiver aperture (number of active elements) is expanded with increasing investigation depth in a process referred to as dynamic aperture. The reason for using this is that the lateral resolution is directly proportional to the ratio of imaging depth to array aperture (commonly referred to as the F-number). To maintain the lateral resolution, more and more elements are included in the aperture as the distance increases, at least up to the point where the number of available channels in the hardware is the limiting factor. The term is used in receive mode, but correspondingly the aperture can also be increased in transmit, but only in a discrete number of steps, resulting in a number of focal zones. This process is however not dynamic as it is for receive, when the aperture opens up for deeper echoes from the same imaging line.

On Figure D.68 the echoes from a shallow reflector A return to the transducer with different time/phase changes between transducer elements when compared with those from a deeper reflector B. The delay and sum instrumentation uses a limited aperture for echoes from A. Echoes from B arrive slightly later and the system uses an increased number of elements with different delays to achieve effective focussing.

Dynamic focusing

(Ultrasound) Dynamic focusing is a technique applied in receive mode to enable a receive focus over the entire length of the interrogated B-mode-line. In Figure D.69, an echo is produced by a scattering object in the interrogated direction. To focus at that depth, delays are introduced to the signals originating from the individual transducer elements. The delay corresponds to the propagation delay, and makes the signals appear to have been received simultaneously. After summation, these signals add constructively, and produce a large signal. Objects out of focus add destructively and are thus suppressed. In dynamic focusing, these delays are adjusted for each point in the interrogated direction so that an optimal focus along the entire line is achieved.

On Figure D.69 as the wavefront from a scattering object approaches the transducer, the wavefront is curved. By applying a delay to the individual signals that corresponds to the expected propagation delay at a given distance, signals arriving from that distance are summed to produce a large echo.

Dynamic imaging

(Nuclear Medicine) Dynamic imaging refers to a set of images acquired in a region of interest over a period of time. For example, dynamic image sets can be used to study the bio-kinetics of tumour targeting agents. By studying the amount of activity in an organ or organ region it is possible to draw conclusions about the uptake, redistribution and clearance of targeting agent in the selected region. The bio-kinetics can then be modelled by a time-activity curve used in dosimetry calculations.

Dynamic jaw collimation

(Radiotherapy) Dynamic collimation is achieved when the beam collimation is changed during the time the beam is delivering treatment. In this case the beam aperture size changes either by movement of a solid jaw or by using the multi-leaf collimator (MLC). The latter is sometimes referred to as dynamic multi-leaf collimation (DMLC).

Related Articles: Collimation, Dynamic wedges, Dynamic multileaf collimation, Asymmetric jaws

Dynamic multileaf collimation

(Readiotherapy) See Multileaf collimation

Dynamic radiography

(Diagnostic Radiology) Dynamic radiography is another name for x-ray fluoroscopy (a method for visualising the organs in motion).

Related Article: Fluoroscopy

Dynamic radiosurgery

(Radiotherapy) Dynamic radiosurgery is a special technique where the gantry moves during the treatment with the beam on. There are two approaches:

Linac-Based Dynamic Radiosurgery: This is radiosurgery on a linac using the arc technique. Often several arcs may be used and couch movement may be used.

Four-Dimensional Dynamic Radiosurgery: This is the use of radiosurgery techniques such as the Cyberknife robotic linac to track target motion in radiosurgery. This is particularly useful for treatment sites which exhibit intra-fraction motion, such as the lung.

Related Articles: Stereotactic radiosurgery, Gamma knife, Robotic linac, Arc therapy, Stereotactic frame

Dynamic range

(General) In signal processing, the term dynamic range usually refers to the ratio of the minimum and maximum of a measurable quantity. In medical imaging, these optima are usually the minimum energy necessary to cause a measurable signal in a detector, and the maximum energy that can be measured without saturation of the signal. In digital imaging devices, this saturation can also be limited by the chosen image depth (i.e. the maximum analogue

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**FIGURE D.67** (See colour insert.) (a) Seven stops, equal dwell times, 10 Gy at 2 cm distance from central stop position. (b) Seven stops, optimised dwell times, 10 Gy at 2 cm distance from all stop positions.
value that can be converted to a given data type). The unit used for quantifying the dynamic range is usually the deciBel (dB), a unit of dimension given in decadic logarithms. The lower bound for converting the energy imparted to the image detector to a signal is usually defined by detector characteristics such as the DQE. Another quantity that defines the lower bound of the dynamic range is bias noise, which is also connected to the signal-to-noise ratio (SNR) of a signal.

**Histograms:** A measure for optimal use of the dynamic range in a medical image is the histogram, which represents the distribution of the measured signal over the dynamic range in an image. An optimum histogram shows a smooth and even distribution with little signal in the low and high end of the dynamic range. If the dynamic range is not fully utilised, or if large parts of the histogram cumulate at the lower or upper bound of the dynamic range, it is likely that the signal resolution is either not fully utilised or that usable signals are not recorded due to saturation or detector inefficacy.

**Transfer Functions:** In general image processing, the transfer function of an input signal \( I_{\text{in}} \) to an output signal \( I_{\text{out}} \), which gives a graphic illustration of the dynamic range in terms of a response curve, is governed by the gamma-value. The gamma-value gives a measure of the slope the transfer function and is used in film-based radiography as a measure of the contrast behaviour of the film used. A high-contrast film usually features a steep gradient; therefore, small input contrast detail differences are recorded over a large span of the dynamic range available. As a consequence, such media is easily subject to over- or under-exposure. In medical imaging, the most important transfer function operation is windowing of images, where a certain part of the histogram known to contain anatomical structures of interest such as soft tissue or the bony skeleton is selected and displayed within a smaller dynamic range; by applying a windowing operation, small contrast detail can be visualised in a manner that is optimal to human perception.

**Modelling the Transfer Function:** A simple mathematical model for a symmetric transfer function can be derived from the logistic function.

It is a sigmoid type of mathematical function given as

\[
I(r) = \frac{D + 1}{1 + e^{-\gamma(r - c)d}}
\]

\(I(r)\) is the resulting signal
\(r\) is the input signal
\(D\) is the numerical range for recording the signal
\(c\) is the window centre (the location of the turning point of the sigmoid)
\(d\) is the width of the window

**Dynamic range**

(Ultrasound) In general, the dynamic range is the ratio between the smallest and largest possible values of a quantity. The dynamic range is usually quoted in decibels, i.e. \(10 \log(P_P/P_0)\) for power related measures, and \(20 \log(A/A_0)\) for amplitude related measures. For instance the dynamic range of a 10 bit A/D-converter can be found as \(20 \log(2^{10})\) which approximately is 60 dB. Ultrasound scanners often quote the dynamic range on-screen which includes effects of multiple elements, time gain compensation (TGC) and non-linear amplification. For instance an aperture using 100 elements, where each channel is using a 10 bit A/D-converter can achieve a dynamic range of 40 + 60 = 100 dB (i.e. the minimum detected signal would be a 1-bit change on one channel, whereas the maximum is the maximum change on all channels). TGC-compression and non-linear amplification (log amplification) can increase this number further.

For ultrasound scanner operators, the term dynamic range is used to describe the on-screen range of echoes displayed. Many scanners permit the operator to alter the displayed dynamic range, e.g. to have a limited dynamic range of echoes, increasing contrast and suppressing weak echoes or alternatively increasing the range to highlight weak echoes. The effect of changing this is shown in Figures D.70 and D.71.

With a reduced dynamic range 40dB (Figure D.70), there is greater contrast within the kidney. However, echoes from the cortex are weak. In the lower image (Figure D.71), the weak echoes from the cortex are now displayed as dynamic range is increased to 65 dB.

**Dynamic spatial reconstructor**

(Diagnostic Radiology) During the early ages (1970s) of CT a major limitation for scanning dynamic objects was the slow speed of gathering information for one image.
The gantry (with x-ray tube and detectors) of the second-generation CT scanners was doing a 360° rotation around the patient for about 1 min, later the third-generation CT scanners minimised this time to several seconds, but still organs such as the heart could not be scanned. This was the reason for new types of CT scanners to be designed – ultra fast CT.

The first such scanner, developed by a group in the Mayo Clinic in United States, was the dynamic spatial reconstructor. This scanner used a gantry with an arc of 28 x-ray tubes. Against each tube is a detector (fluorescent type with photomultiplier). The tubes cover about 180 angular degrees around the patient. This way a 180° scan can be made without any movement, but only by consecutively switching the x-ray tubes.

Additionally the whole gantry rotates slowly around the patient (about 15 revolutions per minute) to cover angles bigger than 180°. During this scanning the x-ray tubes fire short x-ray pulses (several ms each). These consecutive x-ray pulses are synchronised with the rotation, thus minimising the effective time for rotation and creating conditions for collecting the necessary scanning data for a very short time (on the order of 15 ms per scan). This system can scan the heart, but was very expensive and difficult to build, so only prototypes were produced. Later another ultra fast CT system was made (electron beam CT, e.g. Imatron), which was more practical, but still has limited use. The development of contemporary cardiac CT scanners allows for 0.33 s scan of the heart.

**Related Article:** Electron beam CT


### Dynamic susceptibility contrast MRI

*Magnetic Resonance* Measurements of microcirculatory parameters of the brain, such as cerebral blood flow (CBF), cerebral blood volume (CBV) and mean transit time (MTT) add significant value to the functional diagnostics of many common diseases. Perfusion MRI is of potential relevance in, e.g. studies of dementia and trauma, assessment of the ischemic penumbra zone in acute ischemic stroke (in combination with diffusion MRI) to facilitate treatment decisions, investigations of intracranial vascular malformation in connection with neurointerventional procedures, preoperative classification and grading of brain tumours and monitoring of various kinds of cancer therapy.

The most common perfusion MRI method in clinical environments is dynamic susceptibility contrast MRI (DSC-MRI). The DSC-MRI technique yields parametric maps of regional CBF, CBV and MTT and is based on the theory of intravascular or non-diffusible tracers. DSC-MRI requires rapid intravenous injection of a gadolinium-based contrast agent, and after injection the tracer concentration in a tissue-feeding artery as well as in the tissue of interest is monitored by rapid imaging during the first passage of the contrast-agent bolus. The arterial concentration-versus-time curve $C_{ar}(t)$ is often referred to as the arterial input function (AIF).

The dynamic imaging (Figure D.72) is typically accomplished by gradient-echo (GE) or spin-echo (SE) echo-planar imaging (EPI) in multi-slice mode. The employed pulse sequence must enable a temporal resolution of approximately 1.5 s, and the echo
time (TE) must be sufficiently long for the contrast agent to cause a significant signal drop at peak concentration. For GE-EPI, the temporary signal loss during the bolus passage is due to the contrast-agent-induced shortening of tissue $T_1^*$ (i.e. a local susceptibility effect). The signal-reduction effects for SE-EPI appear to be somewhat more complex, including $T_1$ shortening of capillary blood in combination with tissue-water diffusion in the local magnetic-field gradients caused by the contrast agent.

The relationship between the image signal $S$ and the measured tracer concentration $C$ is based on the assumption that $C$ is proportional to the change in the transversal relaxation rate, i.e. $C = k \Delta R^2$. Consequently, $C$ is calculated as $C(t) = (-k/TE) \ln[S(t)/S_0]$, where $i$ is time and $S_0$ is the baseline signal. In order to obtain whole-blood volume in units of mL/100 g, a correction factor $k_H$ is applied to account for brain density and haematocrit differences between large and small vessels, i.e. $k_H = 100[1 - H_{large}]/[p(1 - H_{small})]$, where $H_{large}$ and $H_{small}$ are the haematocrit values in large and small vessels, respectively, and $p$ is the brain density. The CBV is given by the time integral of $k_HC(t)$, normalised to the time integral of the measured arterial concentration $C_{art}(t)$, i.e.

$$CBV = k_H \int_0^{\infty} \frac{C(t)dt}{C_{art}(t)dt}$$

The measured tissue concentration of tracer (corrected by the factor $k_H$) is given by the convolution of $CBF \cdot R(t)$ with the AIF. $R(t)$ is the impulse tissue residue function, i.e. the fraction of tracer still present in the capillary system at time $t$ after an assumed instantaneous input of tracer, i.e.

$$k_HC(t) = CBF \left[ R(t) \otimes C_{art}(t) \right]$$

CBF can thus be obtained by deconvolution of the tissue concentration curve with the AIF. Finally, the mean transit time is given by $MTT = \text{CBV}/\text{CBF}$, according to the central volume theorem. According to Zierler’s area-to-height relation, $MTT$ can also be calculated as $MTT = [R(0)/dt]R_{max}$, where $R_{max}$ is the peak value of the $R(t)$ function retrieved by deconvolution. Examples of parametric perfusion maps from a normal volunteer are given in Figure D.73.

Although DSC-MRI has become a useful tool for assessment of focal perfusion deficits, quantification is hampered by, e.g. blood-brain barrier leakage, limited deconvolution performance in the presence of noise, arterial partial-volume effects, arterial signal saturation at peak concentration, arterial signal displacement due to local geometric distortion at peak concentration and by the fact that contrast-agent relaxivity appears to vary with vessel size and geometry. An additional inherent problem is that arterial dispersion prohibits one single AIF to accurately represent the entire brain, i.e. the deconvolution should ideally employ the AIF of the artery that actually feeds each local capillary segment.

**Related Articles:** Perfusion imaging, Arterial input function, Mean transit time, Cerebral blood flow, Cerebral blood volume


**Dynamic wedge** (Radiotherapy) A dynamic wedge is a technique used to create an effective wedged field using independent collimator jaws. This is achieved by moving one jaw across the field during beam on, whilst keeping the other stationary, so that different parts of the field are irradiated for different lengths of time. The advantage of this method is that the beam does not undergo any beam hardening that would otherwise be associated with a physically wedged field. There is also no need for physical placement of heavy wedges within the path of the beam.

A wedged field of any desired angle can be created by combining this dynamic field with an open field of varying duration. Segmented treatment tables, STTs, are used to define the exact collimator position as a function of cumulated beam monitor units, as needed to create each wedged field. A typical STT is illustrated graphically in Figure D.74. It is usual for the upper collimator jaws to be used for this purpose as they are closer to the source and hence do not need to be physically moved as great a distance than the lower jaws. The length of treatment is restricted by the maximum speed of movement of the collimator jaws.

![FIGURE D.73](image) Maps of cerebral perfusion parameters in two slices in a normal volunteer. From left to right: Colour-coded regional cerebral blood volume (rCBV), colour-coded regional cerebral blood flow (rCBF) and mean transit time (MTT).

![FIGURE D.74](image) A typical STT to create a wedged field. An open field is delivered for the first 20% MU, after which one jaw starts moving continuously across the field, stopping 5 mm away from the stationary diaphragm.
Clinical implementation of dynamic wedges requires measurement of central axis PDDs, central axis wedge transmission factors and transverse beam profiles of the dynamic wedges. These measurements are complicated by the modulation of the photon fluence during the delivery of the radiation field.

The central axis PDD may be measured by integrating the dose at each point during the entire irradiation of the dynamic wedge field.

The central axis wedge transmission factors are determined by taking the ratio of the collected ionisation at a specified depth for the dynamic wedge field to the collected ionisation at the same specified depth for the open field with the same collimator and MU settings. It is important to note that the central axis wedge transmission factors for dynamic wedges may have much larger field size dependence than physical wedges and the field size dependence for dynamic wedges may not be asymptotic. Some manufacturers’ implementations of the dynamic wedge technique show a significant change in the trend of the central axis wedge transmission factor as the field width changes between 9.5 and 10 cm. This change in the central axis wedge transmission ratio has been demonstrated to approach 20%. This characteristic should be carefully investigated on each machine.

Dynamic wedge transverse beam profiles can be measured with a detector array or an integrating dosimeter such as radiochromic film. When a detector array is used, the sensitivity of each detector must be determined.

Dynamic wedges are offered for Varian linear accelerators (enhanced dynamic wedge, [EDW]), and Siemens linear accelerators (virtual wedge, [VW]).

**Abbreviations:** EDW = Enhanced dynamic wedge, PDD = Percentage depth dose, STT = Segment treatment table and VW = Virtual wedge.

**Related Articles:** Wedge field, Dynamic jaw collimation, Asymmetric jaws


**Dynode**

(General) Dynodes are a part of the photomultiplier tube (PM tube) signal enhancer system. A set of dynodes (9–12) are located in a vacuum tube following the photocathode. Photoelectrons are released from interactions in the photo-emissive substance on the entrance window. The photoelectrons are accelerated by the electric field and focused on the first dynode using a focusing grid. The dynode is coated with a substance with similar properties to the photo-emissive substance used in the photocathode. When a high-speed photoelectron ‘hits’ the dynode several secondary electrons are released. The number of electrons released depends on the energy of the photoelectron, i.e. proportional to the voltage difference between the photocathode and the dynode. The electron multiplication process is repeated when the secondary electrons are accelerated against the second dynode which is maintained at a higher potential than the first dynode. The process is repeated for all dynodes leading to a ‘shower’ of electrons reaching the anode. The total signal multiplication factor depends on individual dynode multiplication factors and the number of dynodes. Relatively low stimuli can therefore be enhanced into a large detectable pulse using a PM tube.

**Related Articles:** Photomultiplier (PM) tube, Photocathode of photomultiplier tube

Ear protection

(Magnetic Resonance) During the image acquisition a significant amount of acoustic noise is produced by vibrations of the gradient coils which are embedded in solid material around the bore of the magnet. The electrical current passing through the gradient coils produces a large force which tends to move the coils and causes vibration, thus generating loud sounds such as tones, knocking or banging noises associated with the frequency and waveform of the applied gradient pulse.

The intensity level or relative level of a noise is its intensity relative to an agreed ‘zero’ intensity. The latter is chosen as the noise intensity $I_0$ at the threshold of hearing which is generally accepted as $10^{-12}$ W/m$^2$ (1 pW/m$^2$) at 1 kHz. Since the ear has a logarithmic response to noise intensity, it is convenient to define the unit of intensity level using a logarithmic scale.

The intensity level of a sound of intensity $I$ (W/m$^2$) measured in bels (B) is then defined by

$$\text{Intensity level} = \log_{10}\left(\frac{I}{I_0}\right) \text{B}$$

The bell is a large unit representing intensity differences in the ratio 10:1 and usually it is more common to use a smaller unit, the decibel (dB), given by

$$\text{Intensity level} = 10 \log_{10}\left(\frac{I}{I_0}\right) \text{dB}$$

where 1 B = 10 dB.

In the following table intensity levels are reported for different sources of noise with the indication of the associated average and peak pressure.

<table>
<thead>
<tr>
<th>Source of Noise</th>
<th>Intensity Level (dB)</th>
<th>Sound Intensity (W/m$^2$)</th>
<th>Average Pressure (Pa)</th>
<th>Peak Pressure (Pa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Threshold of hearing at 1 kHz</td>
<td>0 (silence)</td>
<td>$10^{-12}$</td>
<td>0.00002</td>
<td>0.00003</td>
</tr>
<tr>
<td>Library</td>
<td>20</td>
<td>$10^{-10}$</td>
<td>0.0002</td>
<td>0.0003</td>
</tr>
<tr>
<td>Background music</td>
<td>40</td>
<td>$10^{-8}$</td>
<td>0.002</td>
<td>0.003</td>
</tr>
<tr>
<td>Speech at 0.6 m</td>
<td>60</td>
<td>$10^{-6}$</td>
<td>0.02</td>
<td>0.03</td>
</tr>
<tr>
<td>Heavy traffic</td>
<td>80</td>
<td>$10^{-4}$</td>
<td>0.2</td>
<td>0.3</td>
</tr>
<tr>
<td>Jet overhead</td>
<td>100</td>
<td>$10^{-2}$</td>
<td>2</td>
<td>3</td>
</tr>
<tr>
<td>Threshold of feeling</td>
<td>120</td>
<td>1</td>
<td>20</td>
<td>30</td>
</tr>
</tbody>
</table>

The decibel level of noise within the gantry of the MR scanner varies from about 82 dB to about 120 dB, and its intensity level depends on the pulse sequence used for the imaging procedure. Recently noise levels have increased considerably with the introduction of new fast spin-echo sequences during which the gradient amplitudes increase and the rise times decrease.

Acoustic noise levels for fast MRI pulse sequences have been measured on MRI systems with field strengths ranging from 0.2 to 3 T. Measured noise levels varied from 82.5 dB(A) for a 0.23 T system to 118.4 dB(A) for a 3 T system.

dB(A) or A-weighted decibels are decibels with the sound pressure scale adjusted to conform to the frequency response of the human ear.

Analysis of the measurements showed that

1. Pulse sequence parameters, particularly FOV and TR, were more influential in determining noise level than field strength
2. The noise level was found to vary along the z-direction with a maximum near the bore entrance
3. In some case there was a significant increase in noise with a volunteer present instead of a test object

Acoustic noises up to 65 dB are acceptable while between 65 and 95 dB the noise is unpleasant but not directly harmful. Above 95 dB hearing damage can occur.

Noise is a complex disturbance having many component frequencies and producing varying effects on exposed individuals. High intensity noise is generally accepted as greater than 85 dB and as the noise level increases, there are a number of harmful results:

- Anxiety, headaches and temporary hearing losses
- Change in hearing acuity and possible damage to the cochlea
- Stimulation of receptors in the skin
- Significant changes in pulse rate
- Vibration of muscles and reduced coordination
- Feeling of fear, annoyance, dissatisfaction
- Nausea, vomiting, dizziness (>30 dB)
- Pain in the middle ear (=140 dB)

Temporary hearing loss has been also reported using conventional pulse sequences. Gradient noise may also interfere with patient communication.

The International Electrotechnical Commission (IEC) advises MR manufacturers to provide a warning if the acoustic noise from a given pulse sequence is likely to exceed 99 dB(A), which is sufficiently removed from 140 dB(A), the level at which damage to the hearing may occur.

An acceptable and inexpensive means for the prevention of hearing loss is the regular use of disposable earplugs for patients undergoing MR examinations. They can decrease the noise intensity by about 30 dB. A more expensive alternative would be ‘antinoise’ or destructive noise apparatus which, not only reduces noise, but also permits a better communication between the operator and the patient during the examination. They consist of devices performing a real-time Fourier analysis of the noise produced by the gradient coils and generating a signal with the same physical characteristics but opposite phase which is added to the original noise. The subsequent cancellation of the repetitive noise permits a reduction of its intensity level of 50%—70%.
MR operators and accompanying family members will suffer the same noise exposure if they remain in the examination room during the image acquisition and therefore they should use the same hearing protection as the patients.

The following table shows the relationship between the noise duration and recommended permissible sound levels for occupational exposures.

<table>
<thead>
<tr>
<th>Noise Duration per Day (h)</th>
<th>Sound Level [dB(A)]</th>
</tr>
</thead>
<tbody>
<tr>
<td>8.0</td>
<td>90</td>
</tr>
<tr>
<td>6.0</td>
<td>92</td>
</tr>
<tr>
<td>4.0</td>
<td>95</td>
</tr>
<tr>
<td>3.0</td>
<td>97</td>
</tr>
<tr>
<td>1.5</td>
<td>100</td>
</tr>
<tr>
<td>1.0</td>
<td>102</td>
</tr>
<tr>
<td>0.5</td>
<td>105</td>
</tr>
<tr>
<td>0.25</td>
<td>115</td>
</tr>
</tbody>
</table>

These noise levels are based upon a long-term exposure to noise. The levels for chronic exposure concern only those MR operators who are constantly working in a noisy environment.


Earth's magnetic field

The Earth's magnetic field is approximately the field of a dipole positioned at the centre of the Earth. The dipole has a declination by approximately 11.3° from the planet’s axis of rotation. The flux density of the magnetic field at the Earth's surface ranges from some less than 30 μT (0.3 G) to over 60 μT (0.6 G) around the magnetic poles. The Earth's field is changing in size and position. The magnetic poles may shift as much as 15 km a year.

Earthing

Earthing is a term used in UK electrical engineering to represent electrical equipment that has direct physical connection to Earth thus providing protection against electrical shock. In the United States the equivalent term is 'grounding'.

Echo enhancing agent

Echo enhancing agents are used as contrast media in ultrasound examinations. These consist of microbubbles, i.e. micron-sized bubbles, stabilised by a shell. Further details are found under Contrast agents.

Echo planar imaging (EPI)

Echo planar imaging (EPI), proposed by Mansfield in 1977, refers to the concept of collecting all phase encoding steps needed to reconstruct an image using one series of echoes after a single RF excitation. Following the RF excitation, the readout gradient (G_r) is rapidly reversed in combination with short phase encoding gradient pulses (G_y). The sequence diagram and corresponding k-space for a standard gradient-echo EPI sequence are shown in Figures E.1 and E.2, respectively. Each echo is generated by the refocusing action of the readout gradients, and the effective echo time (TE) is defined as the time from the RF pulse to the central echo (k_y = 0).

EPI can also be employed as a spin-echo experiment by adding a 180°-pulse after the 90°-pulse shown in the following. Another variation of the EPI sequence is spiral imaging with two oscillating gradients, generating a spiral k-space trajectory.

EPI is very demanding on the imaging hardware because large field gradients have to be generated and switched rapidly. Another drawback is its sensitivity to magnetic field inhomogeneities present in boundaries between tissue and air. This behaviour arises from the low bandwidth in phase encoding direction and the rather long effective TE, leading to distortion and in severe cases to gross signal dropout.

The speed at which images can be obtained with EPI has allowed several applications for dynamic studies of the brain such as functional MRI (fMRI), diffusion imaging and perfusion imaging.

Related Articles: Diffusion imaging, Echo spacing, fMRI, Gradient echo, Perfusion imaging, Spin echo

Echo-planar imaging and signal targeting with alternating radiofrequency (EPISTAR)

(Magnetic Resonance) The EPISTAR arterial spin labelling (ASL) preparation method is based on the subtraction of two images, one acquired with a preparatory inversion of inflowing arterial spins (the labelling experiment) and one without inversion (the control experiment). For the inversion of arterial spins, a spin tagging radio-frequency (RF) pulse (normally a 180° RF pulse) is applied in an axial slab proximal (caudal) to the imaging plane. In the control experiment, an inversion pulse is applied to a slab distal (superior) to the imaging slice in order to cancel magnetisation transfer effects. After a time delay TI the image is acquired with a fast read-out sequence such as EPI, spiral reconstruction, HASTE or 3D GRASE. One or more presaturation pulses are applied immediately before the inversion pulse to suppress signal from the static tissue. The inverted arterial spins, in labelling experiment, cause a degeneration of the MR signal in the imaging slab. The degree of the MR signal depression corresponds to the amount of the regional perfusion and the subtracted image can be viewed as a qualitative perfusion-weighted image. By fitting the subtracted image to a model, usually used is the general kinetic model developed by Buxton et al. (1998) a quantitative perfusion maps can be obtained. The major disadvantages of this method are: the compensation for magnetisation transfer effects is complete only half way between the inversion slabs which makes method suited only for single slice imaging and the blood, inverted by a control pulse, enters into the imaging slice from the distal side and affects the subtracted image negatively (Figure E.3).

Related Articles: Perfusion imaging, Arterial spin labelling, FAIR, PICORE, QUIPSS – QUIPSS II – Q2TIPS


Echo ranging

(Ultrasound) Echo ranging is a distance measurement performed by measuring the time interval between the transmission of a sound pulse and the return of its echo. See also Pulse echo.

Echo spacing

(Magnetic Resonance) Echo spacing refers to the distance in time between the echoes in pulse sequences employing multiple echoes, e.g. echo planar imaging and fast spin echo. The echo spacing affects the amount of image artefacts and the maximum number of slices per repetition time.

Related Articles: Echo planar imaging, Fast spin echo

Echo time (TE)

(Magnetic Resonance) Echo time, TE, is the time from the RF excitation to the centre of the echo being received. Together with repetition time (TR) and excitation pulse flip angle, TE is an important parameter in the build-up of image contrast. TE determines how much decay of the transverse magnetisation is allowed to occur before the signal is read. Short TE allows less T2 or T2* signal decay while a longer TE leads to a smaller signal. A long TE in combination with a long TR produces T2-weighted images. A short TE together with short TR leads to T1-weighted images. A proton-weighted image is obtained when a short TE in combination with a long TR is used (Figure E.4). TE is often given in milliseconds.

Related Articles: Effective echo time, Flip angle, Repetition time, TR, T1-weighted, T2-weighted

Echo train length (ETL)

(Magnetic Resonance) Echo train length is the number of echoes created after one excitation. In particular, this refers to the number of applied 180° rephasing pulses and their corresponding echoes during one repetition time (TR) for a fast spin echo pulse sequence type. Another name for echo train length is turbo factor. See Fast spin echo (FSE) for more information.

Related Articles: Fast spin echo (FSE), Half acquisition single-shot turbo spin echo (HASTE), Rapid acquisition relaxation enhancement (RARE), Turbo spin echo (TSE)

FIGURE E.3 The principle of an EPISTAR pulsed ASL labelling technique. (a) Illustrates the labelling experiment and (b) shows the corresponding control experiment.

FIGURE E.4 Spin-echo pulse sequence. RF and ADC are the radio frequency pulse and the signal received from the slice. GS, GP and GF are the slice selective, phase and frequency encoding gradients. TE is the time between the 90 pulse and the ADC.
Echocardiography refers to ultrasound examination of the heart. The echocardiogram gives information as to the structure and motion of the heart and the blood flow within it. A typical cardiac examination may use many of ultrasound’s modalities; M-mode (Figure E.5), B-mode, colour flow imaging and pulsed and continuous wave spectral Doppler. Recent developments in echocardiography include 3 and 4D imaging, contrast agent imaging and tissue Doppler imaging. The examination includes the pericardium (the tissue surrounding the heart), the structure of the heart including the chambers and valves and the velocity of blood flow in the heart.

Echocardiography is used to examine many cardiac conditions including heart valve disease, ischaemic heart disease, cardiomyopathies, pericardial disease and cardiac masses. It is used to examine the great vessels, in pulmonary disease, in arrhythmias and in the assessment of left ventricular function.

The approach may be transthoracic which is performed through the chest wall (Figure E.6) or transesophageal where the transducer is passed endoscopically into the esophagus. The examination can be combined with exercise – a stress echocardiogram.

Related Article: Stress echocardiography

Echogenic (Ultrasound) Echogenic, or hyperechoic, describes echoes which are unusually bright. An example is shown in Figure E.7 where the cortex and medulla appear markedly brighter than is usually seen. The kidney can be described as echogenic.

Eddy currents (Magnetic Resonance) Electric currents can be induced in a conductor due to changes in magnetic field or by motion of the conductor through a magnetic field. These electric currents are called eddy currents. When the magnetic field is changed it leads to changes in flux, $\Phi$, and according to Faraday’s law of induction, the change in flux through an area, $\Omega$, creates an electromotive force (emf) in the conducting material:

$$\text{emf} = -\frac{\partial \Phi}{\partial t} = \int_{\Omega} \frac{\partial B}{\partial t} dA$$

where

- $B$ is the magnetic field
- $dA$ is a small part of the area $\Omega$

The main source of eddy currents is the rapidly varying gradient fields. When the gradients are switched on and off, the change in magnetic field induces a current in the conductor in the opposite direction to counteract the change in magnetic field (see Figure E.8).

In the MR environment the eddy currents are induced by the gradient fields in all conducting material in the surrounding environment (e.g. the cryostat). The currents can introduce artefacts to the images and seriously degrade overall magnetic performance. Shielding of gradient coils as well as inclusion of eddy current compensation in the gradient pulse design are frequently used methods to reduce eddy current effects (Figure E.9).

In the right image strong diffusion gradients ($b = 800 \text{s/mm}^2$) result in distortions.

Edge detection (Nuclear Medicine) Edge detection is the process of identifying organ or organ substructure boundaries in morphological images.
Edge enhancement

(General) In image processing it is sometimes of importance to enhance local changes in count density to visualise edges or boundaries. For example, the volume of an organ may need to be calculated and in order to do this the boundary of the organ needs to be determined. It is necessary to use an imaging modality that shows a contrast difference between the organ of interest and the background. The method identifies those pixels that have largest differences in pixel values from their local neighbours. For example, consider the following mask:

\[
\begin{array}{ccc}
  -1 & -1 & -1 \\
  -1 & 8 & -1 \\
  -1 & -1 & -1 \\
\end{array}
\]

When convolving an image with this mask, the results will be zero for all pixels whose neighbours have the same value but differ from zero if there is a local difference in pixel values. An edge in an image will then be intensified when applying this filter. The filter can be modified to enhance edges in a particular direction. Examples are the Prewitt operator:

\[
\begin{array}{cccc}
  -1 & -1 & -1 & 0 & -1 \\
  0 & 0 & 0 & -1 & 0 \\
  -1 & -1 & -1 & 0 & -1 \\
\end{array}
\]

Edge spread

(Nuclear Medicine) The edge-spread function (ESF) is closely related to the point-spread function (PSF) and line-spread function. An edge is imaged and the spatial resolution and sharpness degradation is evaluated by defining a profile through it. The ESF is then used to calculate the modulation transfer function (MTF) which is a measure that describes the degradation of an imaging system in terms of frequencies and amplitudes.

Abbreviations: ESF = Edge-spread function, LSF = Line-spread function and PSF = Point-spread function.

Related Article: Modulation transfer function

Editing, spectral

(Magnetic Resonance) Spectral editing refers to a range of techniques employed to simplify NMR spectra and so facilitate interpretation and quantification, particularly when the appearance of spectra is complicated by overlapping lines.

Water suppression is a common example of spectral editing. More generally, the following are examples of spectral editing technique.

1. Relaxation-based techniques: When a spectrum contains one or more overlapping lines with different $T_1$ relaxation times, a broadband inversion pulse followed by a suitable delay can be used to reduce (or ideally null) signal from one of the overlapping lines. Similarly, a spin-echo experiment can be used when the $T_2$ relaxation times of the two species to the extent that it is possible to chose an echo time such that signal from one species remains while that from the other has substantially decayed.

2. $J$-coupling-based techniques: These methods can be employed when one of the overlapping species is a coupled nucleus. In such a case, the signal from this species is modulated in amplitude and phase during the echo time of a spin-echo experiment. This phenomenon can be exploited by performing a series of experiments in which the echo time is changed, and in some cases also using decoupling, followed by signal subtraction to eliminate unwanted lines and enhance others. Various polarisation transfer and multiple quantum filtering techniques, beyond the scope of this article, may also be considered examples of techniques in this category.

3. Two-dimensional NMR: This is a very general class of technique used in chemical analysis, the details of which are beyond the scope of this article. Briefly, a 2D NMR experiment consists of a series of acquisitions in which spins are allowed to evolve for an incremented period of time under the influence of some process or interaction.
Effective dose ($E$)

(Radiation Protection) This is the radiation dose quantity in radiation protection used for purposes of risk comparison. It provides an estimate of the dose to the whole body from a single or series of partial body irradiations.

The starting point for determining effective dose is the absorbed dose. Absorbed dose modified by radiation weighting factor, $w_R$, produces the equivalent dose, $H$. Effective dose is a summation of the equivalent doses to each of the organs irradiated modified by their tissue weighting factor, $w_T$. Thus

$$E = \sum w_T H_T$$

The unit of effective dose is the Sievert (named after Rolf Sievert). Tissue weighting factors are loosely based on tissue sensitivity to ionising radiation and have been defined by ICRU (International Commission on Radiological Protection). The current values for tissue weighting factors determined by ICRP in 2005 are given in the following table:

<table>
<thead>
<tr>
<th>Tissue</th>
<th>$w_T$</th>
<th>$\sum w_T$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bone marrow, breast, colon, lung, stomach</td>
<td>0.12</td>
<td>0.60</td>
</tr>
<tr>
<td>Bladder, oesophagus, gonads, liver, thyroid</td>
<td>0.05</td>
<td>0.25</td>
</tr>
<tr>
<td>Bone surface, brain, kidneys, salivary glands, skin</td>
<td>0.01</td>
<td>0.05</td>
</tr>
<tr>
<td>Remainder tissues</td>
<td>0.10</td>
<td>0.10</td>
</tr>
</tbody>
</table>

To indicate how the effective dose would be calculated, consider a procedure in which an irradiation of the thorax delivered absorbed doses of 15 mGy to the thyroid, 10 mGy to the lung, 30 mGy to the breast, 45 mGy to the heart and 10 mGy to the skin. As the absorbed dose was due to x-radiation the radiation factor, $w_R$, is 1 and so the equivalent dose to each organ will be numerically equal to the absorbed dose. Then using the factors in the preceding table, the effective dose can be calculated as follows:

$$E = 15 \times 0.05 + 10 \times 0.12 + 30 \times 0.12 + 45 \times 0.1 + 10 \times 0.1$$

$$= 0.75 + 1.2 + 3.6 + 4.5 + 1$$

$$= 11.05$$

$$E = 11.05 \text{ mSv}$$

Note that the dose to a particular organ, e.g. the lung, is the dose averaged over the whole organ. Partial irradiation of organs is not considered in the calculation of effective dose.

Related Articles: Absorbed dose, Equivalent dose, Radiation weighting factors, Tissue weighting factors

Effective dose equivalent

(Radiation Protection) This is the term that was formerly used to describe the effective whole body dose equivalent to a partial body exposure. This term is no longer used.

The current term is Effective dose.

Related Article: Tissue weighting factor

Effective dynamic range

(Magnetic Resonance) See Dynamic range

Effective echo time

(Magnetic Resonance) The effective echo time, $TE_{eff}$, is the time period between the excitation pulse and the time when the central $k$-space line, corresponding to the echo without phase encoding, is acquired. See Fast spin echo (FSE) or Echo planar imaging (EPI) for more information.

Related Articles: Echo train length, Fast spin echo (FSE), Echo planar imaging, Half acquisition single-shot turbo spin echo (HASTE), Turbo spin echo (TSE)

Effective energy

(General) Effective energy is a characteristic of a polyenergetic x-ray beam that compares it to a monoenergetic beam based on penetrating capability. The effective energy of an x-ray beam can be determined by measuring the HVL, and from that calculating the effective linear attenuation coefficient $\mu_{eff}$. The effective energy of the beam is the photon energy that has the same attenuation coefficient value as determined from published references. In practice the effective energy of a moderately filtered x-ray beam is approximately 2/3 of the maximum photon energy (peak kV). For example, an x-ray beam produced with 90 kVp will have an effective energy of approximately 60 kV.


Effective focal spot

(Diagnostic Radiology) See Focal spot, effective

Effective half-life

(Nuclear Medicine) Effective half-life is the time period in which the amount of deposited radionuclides in an organism is reduced to 50% of the initial value. The decrease in radionuclide is due to two different processes – physical decay and biological excretion. The effective half-life $T_{eff}$ depends on the physical half-life of the radioisotope $T_{phys}$, and biological excretion $T_{bio}$:

$$T_{eff} = \frac{T_{phys} \cdot T_{bio}}{T_{phys} + T_{bio}}$$

The effective half-life can also be derived from the physical and biological decay constants $\lambda_{phys}$ and $\lambda_{bio}$ respectively. The effective decay constant $\lambda_{eff}$ is the sum of the physical and biological decay constants:

$$\lambda_{eff} = \lambda_{phys} + \lambda_{bio}$$

Effective point of measurement

(Radiotherapy) The dose is the expectation value of the energy imparted per unit of mass of an infinitesimal volume centred in a specific point. The dose takes a value at each point of an irradiated medium and describes the spatial distributions of the
Effective source point

Electron beams appear to originate from a point in space that does not coincide with the scattering foil or the accelerator exit window. The term effective (or apparent or virtual) source point (or position) was introduced to indicate the virtual location of the electron source.

Effective source point is relevant to electron beams since they have passed through a scattering filter. This will change the beam from its well defined collimated shape to one which diverges. The degree of divergence will be energy-dependent and will mean that electron beams with different energies will appear to have originated at different source positions.

Source position is an important factor when calculating the change in output factor for extended SSD treatments.

**Abbreviation:** SSD = Source to surface distance.

**Related Article:** Apparent source position


Effective SSD

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Source position is an important factor when calculating the change in output factor for extended SSD treatments.

**Abbreviation:** SSD = Source to surface distance.

**Related Articles:** Apparent source position, Effective source point, apparent focal spot, Electron oblique incidence, Oblique incidence


Effective voltage value

*General* Effective voltage value is the voltage value that has the same effect and gives the correct result on a power calculation as does a DC voltage of the same value. Effective voltage is equal to the square root of the mean value of the squares of the magnitudes of an AC voltage measured at each instant over a defined period of time, usually one cycle. The effective voltage is also known as the RMS (root mean square) voltage (see the eponymous article).

**Related Article:** RMS (root mean square) voltage

**EFOMP**

*EFOMP:* The European Federation of Organisations for Medical Physics, EFOMP, was founded in 1980. Currently (2009) the federation consists of 34 national member organisation and three affiliated national organisations, representing more than 5000 physicists and engineers working in the field of medical physics.
Since its inauguration the main objective of EFOMP has been to harmonise and promote the best practice of medical physics in Europe. In order to accomplish this goal, EFOMP has presented a number of unanimously adopted policy statements, making recommendations on the appropriate general responsibilities and roles of the medical physicist and proposing guidelines for education, training and accreditation programs in medical physics. The aims and purposes of EFOMP also include collaboration with national and international organisations and encouraging exchange and dissemination of professional and scientific information.

**International Collaboration:** EFOMP is part of a larger international medical physics network through the International Organization for Medical Physics, IOMP, where EFOMP is an affiliated regional organisation. IOMP, together with the International Federation of for Medical and Biological Engineering, IFMBE, are the constituent organisations of the International Union for Physical and Engineering Sciences in Medicine, IUPESM. IUPESM is an international network of physical scientists and engineers dedicated to improving health care and well-being world-wide, especially in developing countries. In its role of an international scientific union, IUPESM is a member of the International Council for Science, ICSU.

**Abbreviations:** IC = International Council for Science, IFMBE = International Federation of Medical and Biological Engineering, IOMP = International Organization for Medical Physics and IUPESM = International Union for Physical and Engineering Sciences in Medicine.

**Hyperlinks:** EFOMP: www.efomp.org; IOMP: www.iomp.org; IFMBE: www.ifmbe.org; IUPESM: www.iupesm.org; ICSU: www.iccsu.org

### Eigenfunctions

(Nuclear Medicine) For the linear operator $L$, an eigenfunction defined on a function space is every nonzero function, $f$, which returns the original value of the function multiplied by a discrete factor $\lambda$ also called eigenfunction:

$$L f = \lambda f \quad (E.1)$$

### Eigenvalues

(Nuclear Medicine) Eigenvalues $\lambda$ are a set of scalars for a specific linear system of equations.


### Elastic scattering

(Radiation Protection) Also known as coherent scattering, or Rayleigh scattering.

When the distance between a charged particle and an atom is less than the diameter of the atom, the particle experiences a Coulomb-force interaction with the nucleus. The interaction may take the form of either elastic scattering or radiation energy loss. These interactions are only important for light charged particles such as electrons and positrons.

Most of these events are elastic scattering, in which the electron is deflected with no significant transfer of energy to the medium. This is the cause of the tortuous path electrons follow through a medium. The likelihood of elastic scattering varies with $Z^2$. Most elastic scattering events at diagnostic x-ray and low gamma ray energies are defined as Rayleigh scattering.

The 2%–3% of events that are inelastic are termed radiation interactions. The incident light charged particle is decelerated and deflected by the electric field of the nucleus and emits a significant proportion (up to 100%) of its energy as a Bremsstrahlung x-ray. The cross section for radiation events varies with the square of the absorber atomic number $Z^2$, as well as $E_k$ and $1/m^2$ where $E_k$ and $m$ are, respectively, kinetic energy and mass of the particle.

**Related Articles:** Coherent scattering, Rayleigh scattering, Inelastic scattering

### Electric arc

(Ultrasonography) Ultrasound elastography is the use of ultrasound to produce measurements or images of tissue stiffness. In order to produce elastography images, a force must be applied to produce a strain. The nature and source of the force used to produce a strain can take very different forms, e.g. a force applied by the operator, an external mechanical device, physiological motion or an acoustic force generated by the ultrasound probe.

It is possible to derive a stiffness image by comparing an image acquired before a stress is applied, with one acquired after. To improve performance, most implementations compare raw ultrasound data across several frames. A hard object will tend to move as a whole, whereas soft tissue compresses more unevenly, with tissues closer to the applied force compressing more than those further away. These features are illustrated by Figure E.1.1.

The simplest method of obtaining data under different stresses is for the operator to provide a force. This can be, e.g. from a slow, steady pressure or a gentle bouncing motion. However, because of the operator dependence, these methods do not provide fully quantitative data of tissue stiffness.

There are several alternative techniques. Sonoelastography uses an externally applied low-frequency vibration as the stress, and makes use of Doppler techniques to derive the strain. Vibroacoustography uses an oscillating acoustic force to apply the stress. A more recent development is that of acoustic radiation force imaging (ARFI). This technique produces a push-pulse of high intensity ultrasound to displace the tissue, and then lower intensity pulses to image. This technique may enable quantitative measurements to be made. Several suppliers are working on bringing ARFI-based systems to market, but at the time of writing none are yet available.

### Electric arc

(Diagnostic Radiology) Electric arc (or voltaic arc) is a phenomenon of electrical current flowing through normally non-conductive medium, driven by high electric voltage.

The gas discharge which occurs in some (old) x-ray tubes is a phenomenon similar to short electric arc (or long electric spark). It
can occur at situations when gas ions are allowed inside the tube envelope (either due to micro cracks in the glass envelope, or gas extraction from the electrodes due to the high vacuum). These ions, being in the field of high voltage between the anode and the cathode, create a large spark (or cluster of sparks) which results in very high anode temperature and anode current (plus bright light) and could destroy the x-ray tube. X-ray generators often have fast fuses and spark distinguishers to prevent such problems.

Usually electric arc is considered to be the strong current of a continuous discharge, while momentary electric discharges are considered electric sparks.

**Electric current**

(General) Electric current is the flow of electric charge carriers. The electric charge carriers may be either electrons or ions. In a wire, electric current is a flow of electrons that have been dislodged from atoms and is a measure of the quantity of electrical charge passing any point of the wire per unit time. In gases and liquids the electric current is flow of positive ions in one direction together with a flow of negative ions in the opposite direction. Conventionally, the direction of electric current is that of the flow of the positive ions. In alternating current (AC) the motion of the charges is periodically reversed. The SI unit of electric current intensity is the ampere, a flow of 1 C of charge per second, or $6.24 \times 10^{18}$ electrons per second.

**Electric dipole**

(General) The electric dipole is the combination of two equal point charges of opposite sign $q$ separated by a distance $a$.

The dipole moment is defined as

$$\vec{p} = qa\hat{a}$$

The electrostatic potential from a dipole at a point $P$ is given by

$$V = \frac{1}{4\pi\varepsilon_0} \left( \frac{q}{r_1} - \frac{q}{r_2} \right) = \frac{1}{4\pi\varepsilon_0} \frac{q(r_2 - r_1)}{r_1 r_2}$$

where $r_1$ and $r_2$ are, respectively, the distance of the positive and negative charge from $P$ (Figure E.12).

When $a \ll r$ the electrostatic potential $V$ becomes

$$V = \frac{1}{4\pi\varepsilon_0} \frac{qa\cos\theta}{r^2} = \frac{1}{4\pi\varepsilon_0} \frac{p\cos\theta}{r^2}$$

**Electric field**

(General) An electric charge or a distribution of charges will cause a force $F$ to act on some other charge located in the surrounding space. The electrical field intensity denoted as $\vec{E}$ at a point in the space is defined as

$$\vec{E} = \frac{\vec{F}}{q}$$

where $q$ is a small charge assumed positive placed at that point. The direction of the electric field is the same as the direction of the force it would exert on a positively-charged particle and
opposite to the direction of the force on a negatively-charged particle. The SI unit of electric field intensity is the Newton per coulomb (N/C). An equivalent unit for the electric field intensity is the volt per meter (V/m). Electric fields follow the superposition principle and therefore, at any point, the total electric field created by more than one charge is equal to the vector sum of the respective electric fields that each charge would create in the absence of the others.

**Electric power**  
*General* The energy dissipated by an electric current per unit of time. It is calculated by multiplying the current consumption by the applied voltage. The SI unit of power is the Watt.

*Related Article:* Watt

**Electrical charge**  
*General* See Charge

**Electrical interlock**  
*Radiotherapy* See Interlock; Interlocking device

**Electrical resistance**  
*General* See Resistance, electrical

**Electricity, static**  
*General* See Static electricity

**Electrocardiographic triggering**  
*General* See Cardiac gating

**Electrode**  
*General* An electrical terminal is used to make contact between metallic and non-metallic parts of an electric circuit. It is most often used to pass a current through an electrolyte, gas, vacuum or semiconductor. Electrodes are usually in the form of rod, plate or a wire. Metal electrodes may be made of a copper, lead, platinum, silver, or zinc. Non-metal electrodes are commonly made of carbon. The electrode through which current passes from the metallic to the non-metallic conductor is called the anode, and that through which current passes from the non-metallic to the metallic conductor, the cathode. Typical electrode applications are electrochemical, electrolytic and voltaic cells, electrolytic capacitors, rechargeable batteries, vacuum tubes and semiconductors.

**Electrolytic capacitor**  
*General* See Capacitor

**Electromagnet**  
*Magnetic Resonance* An electromagnet produces a magnetic field by an electric current through the conductor geometry. The simplest design is the solenoid (Figure E.15) producing a homogeneous field within the ‘coil’ and a dipole field at a distance. By placing ferromagnetic material (commonly soft iron) inside the coil the outside magnetic field is enhanced. The ‘toroid’ design is a closed loop solenoid to produce a strong homogeneous field in the gap between the terminal shoes.

The design of superconducting MR-scanners or spectrometers is commonly solenoidal. The open design of resistive MR-scanners usually features two coils in either a horizontal ‘double donut’ (comparable to a solenoid) or a vertical C-shape configuration (comparable to a toroid) (Figure E.16). The former design features air cores and static fields of maximum flux density $B_0 = 0.2$ T, the latter 0.6 T with an iron core.

**FIGURE E.15** A conductor is winded around a core of a para- or ferromagnetic material. When a current, $I$, is passed through the conductor a magnetic field is generated. The polarity depends on the direction of the current.

**FIGURE E.16** An MRI-scanner with an electromagnet is built by two magnets that create the static magnetic field, $B_0$, in-between. When the current, $I$, is switched on, the magnetic field is created.

**Related Articles:** Magnet, Permanent magnet, Resistive magnet, Superconductive magnets

**Electromagnetic energy spectrum**  
*General* The electromagnetic spectrum is a collection of different electromagnetic wave types. Electromagnetic waves are energy waves that are propagated by paired, transversely oscillating electric and magnetic fields and can be classified as one of the following types – radiowaves, microwaves, infrared, visible, ultraviolet, x-rays or gamma rays.

These wave types are defined by different wavelengths, frequencies or energies (Figure E.17). The spectrum can be divided into two unequal parts – ionising radiation and non-ionising radiation. Most types of wave in the spectrum are employed across the different areas of medical physics and clinical engineering. There are no clear boundaries between the different wave types and they tend to overlap with their neighbours. The different frequencies and wavelengths dictate how the waves will interact with matter.

**Related Articles:** Radiation, Ionising radiation, Non-ionising radiation, Ultraviolet radiation, Infrared radiation, Microwaves

**Electromagnetic field**  
*Magnetic Resonance* Electromagnetic fields are produced by electrically charged particles and can be described as either a wave or a particle with zero mass travelling through empty space with the
Electromagnetic hyperthermia

The speed of light. As an example, an antenna functions by oscillations of charged particles along an electrically conductive structure. This causes the emission of an electromagnetic wave. The description of the electromagnetic fields as waves is then suitable and generally applies at the lower range of frequencies (longer wavelength), e.g. for radiowaves and microwaves. Although both descriptions would be equally valid, at higher frequencies (e.g. x-rays and gamma rays) a more suitable model will be to describe electromagnetic fields as particles with a specific energy and zero mass, denoted photons.

Electromagnetic fields can also be described separately as an electrical field and a magnetic field. Stationary charged particles can then be considered the source of the electrical field and moving charged particles (current) can be considered the source of the magnetic field.

The electromagnetic fields can be described mathematically by Maxwell’s equations. For details about these very fundamental relationships, please refer to any suitable educational physics textbook.

In MRI three types of electromagnetic fields are encountered:

1. The static magnetic field, which, e.g. interacts with moving charged particles and objects.
2. The switching magnetic gradient fields (in the order of 1 kHz) which interact with, e.g. the patient’s body and cause induced currents, potentially leading to nerve stimulation.
3. The radiofrequency field (42.6 MHz/T) which interacts with the nuclei at resonance but which also causes heating through induced current and interaction with dipoles in the body.

In order to limit adverse effects of electromagnetic radiation in MRI legal limits apply.

**Related Articles:** Radiofrequency, Radiowaves, Eddy currents, Gradient field

**Electromagnetic hyperthermia**

*(Radiotherapy)* Hyperthermia involves the use of heat, usually in the temperature range 37°C–50°C, as a technique for the treatment of cancer patients. Studies in tissue cultures have shown that animal and human tumour cells are sensitive to temperature changes of a few degrees and that tumours can regress following heating. In addition, heating can raise the radio-sensitivity of tumour cells and reduce their ability to repair radiation damage. Because of this hyperthermia has been employed with radiotherapy, and more recently with chemotherapy, for the treatment of cancer.

Ultrasound as well as electromagnetic radiation (electromagnetic hyperthermia) such as radio-frequency electric fields and microwaves have all been used for hyperthermia. It is important to note that temperature has to be carefully controlled since changes of as little as 0.5°C can lead to significant changes in biological effect.

There are three main types of hyperthermia:

1. Local hyperthermia heats a very small area, usually the tumour itself. Depending on the location of the tumour, the heat may be applied to the surface of the body (superficial hyperthermia), inside normal body cavities, or deep in tissue through the use of needles or probes (interstitial hyperthermia).
2. Regional hyperthermia heats a larger part of the body, such as an entire organ or limb. Usually, the goal is to weaken cancer cells so that they are more likely to be killed by radiation and chemotherapeutic medications.
3. Whole-body hyperthermia heats the entire body. It is typically used to treat metastatic cancer.

Because of their varying thermal characteristics, achieving and maintaining optimal temperature in different tissues remains a problem. One important factor is the rate of cooling which is related largely to blood flow.

**Related Article:** High-intensity focused ultrasound (HIFU)


**Electromagnetic spectrum**

*(General)* See Electromagnetic energy spectrum
Electrometer

(Radiotherapy) The charge produced in an ionisation chamber (i.c.) under typical irradiation condition is small. An exposure rate of $2.58 \times 10^{-4} \text{C/kg min}$ produces in $1 \text{cm}^3$ volume containing approximately $1.29 \times 10^{-6} \text{kg}$ air at standard conditions an ionisation current of $5 \times 10^{-12} \text{A}$. Depending on the ionisation chamber volume and the exposure rate the currents range from about $10^{-12}$ to $10^{-10} \text{ A}$ below the typical detection limit of a galvanometer which is of the order of $10^{-8} \text{A}$. These small current may be measured connecting the i.c. in series with a sensitive current device or connecting the i.c. to a charge measuring device. In the former case the current is measured during the irradiation while in the latter the charge accumulated during an interval of time is measured. The charge measuring device is called electrometer. An electrometer used in conjunction with an ionisation chamber is a high gain, negative feedback, operational amplifier with a standard resistor or a standard capacitor in the feedback path to measure the chamber current or charge collected over a fixed time interval. In Figure E.18 the circuit diagram of an electrometer used in integrating mode is shown.

Several requirements are needed for an acceptable performance of the electrometer:

- Zero stability with fast warm up and long-term drift
- Linearity
- Noise level
- Input bias and offset requirements
- Speed of response and slew rate
- Ruggedness

Electron

(General) The electron is a negatively charged subatomic particle (Table E.1). It has been discovered in 1897 by Joseph John Thomson. In the classical model of the atom, electrons exist in orbits around the positively charged nucleus of the atom. The number of electrons in an uncharged (neutral) atom equals the number of protons in the nucleus, given by the atomic number of the atom.

Electrons occupy shells in the atom defined as K, L, M, etc. shells moving out from the nucleus of the atom. Electrons closest to the nucleus have the highest electronic binding energies.

More correctly, in the quantum model, electrons exist in zones of probability around the nucleus called orbitals rather than precise orbits. The K, L and M correspond to principal quantum numbers of 1, 2, 3. The full set of quantum numbers is given by the principal, orbital, magnetic and spin quantum numbers. Under the Pauli exclusion principle no two electrons in an atom can have the same set of quantum numbers. This principle underlies the electronic structure of the elements and determines how many electrons can occupy a given electronic ‘shell’.

**FIGURE E.18** Circuit diagram of an electrometer used in charge measurement mode.

**TABLE E.1** Properties of the Electron

<table>
<thead>
<tr>
<th>Property</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Charge</td>
<td>$-1.602 \times 10^{-19} \text{C}$</td>
</tr>
<tr>
<td>Rest mass</td>
<td>$0.11 \times 10^{-31} \text{kg}$</td>
</tr>
<tr>
<td>Rest mass energy</td>
<td>$0.511 \text{MeV}$</td>
</tr>
</tbody>
</table>

**FIGURE E.19** Schematic diagram of Compton effect. (Drawing courtesy of E. Podgorsak.)

Electron angular distribution

(General) The recoil electron angular distribution in Compton effect is a plot of the probability of scattering versus angle for an electron involved in a Coulomb interaction event. The distribution will be a function of the photon energy and the scattering medium. The following plot shows an approximate angular distribution for emission of Compton electrons generated by a 1 MeV incident photon (Figures E.19 and E.20).

The photon scattering angle $\theta$ and the recoil electron angle $\phi$ are related through the following relationship:

$$\cot \phi = (1 + \epsilon) \tan \frac{\theta}{2}$$

where $\epsilon$ is the incident photon energy $h\nu$ normalised to the rest mass energy of the electron $m_c^2 = 0.511 \text{MeV}$.


Electron angular scattering power

(General) The electron angular scattering power is a measure of the electron angular scattering distribution. The ICRU defines the mass angular scattering power $T/\rho$ as the rate of change of the mean square scattering angle $d\theta^2$ per unit mass thickness $\rho$ traversed:

$$T = \frac{\rho}{\rho^2} \int d\theta^2$$

and provides the following expression for calculating $T/\rho$:

$$\frac{T}{\rho} = 4\pi \frac{N_A}{A} \epsilon_{c Z}^2 Z (Z + 1) \left[ \frac{1 + \tau}{\tau(2 + \tau)} \right] \ln \left( \frac{1 + \theta_{\text{max}}^2}{\theta_{\text{min}}^2} \right) - 1 + \left[ \frac{1 + \theta_{\text{max}}^2}{\theta_{\text{min}}^2} \right]$$
where

\[ \theta_{\text{max}} = \frac{2A^{1/3}}{\alpha \beta (\tau + 1)} \]

is the cut-off angle due to the finite size of the nucleus given by the ratio of the reduced de Broglie wavelength of the electron to the nuclear radius, and

\[ \theta_{\text{min}} = 1.13 \frac{\alpha Z^{1/3}}{\beta (\tau + 1)} \]

is the screening angle due to the screening of the nucleus by the atomic orbital electrons with

- \( \alpha \), the fine structure constant (1/137)
- \( \beta \), velocity of electron normalised to speed of light in vacuum
- \( \tau \), ratio of electron kinetic energy to its rest energy
- \( A \), nucleon number
- \( Z \), atomic number


**Electron applicator** *(Radiotherapy)* In order to provide a usable electron treatment beam, it is necessary to attach an electron applicator (sometimes called cone) to the head of the linear accelerator. These applicators typically come in a range of set field sizes (e.g. 6 × 6, 10 × 10, 15 × 15, 20 × 20 and 25 × 25 cm²).

The applicator is needed because the penumbra produced without it would be clinically unacceptable. This is due to the fact that while some beam shaping is provided by the secondary collimators in the head of the linear accelerator there is a significant amount of scatter both within the linear accelerator and in the air between the accelerator and the patient. Therefore the applicator collimates the beam and defines it typically at a distance of 5 cm from the patient. Some applicators are also used to provide additional electron scatter thus improving the flatness of the beam.

If fields other than the sizes produced by the applicator set are required, it is common to create an appropriately shaped alloy that can be inserted into the end of the applicator.

**Related Article:** Collimation

**Electron arc aperture** *(Radiotherapy)* Electron treatment can be delivered by static fields with no movement of the gantry. It can also be delivered with the gantry in motion in order to treat an extended area, e.g. the chest wall. This is called electron arc therapy.

Electron arc aperture is the special beam applicator or tray (insertion) which is inserted into the beam limiting system to enable electron arc therapy.

**Related Articles:** Electron beam, Arc therapy, Curvature correction

**Electron arc therapy** *(Radiotherapy)* Electron arc therapy is a form of electron beam radiotherapy delivered by linear accelerator. The treatment technique is designed for treatment of large area underneath and across the chest wall or a superficial target volume involving a large area on or near the body surface. A specially designed electron beam collimator shutter instead of an electron treatment applicator is not used for treatment delivery. The collimator shutter produces an elongated but narrow strip of radiation field of about 2 cm wide and a length that matches the length of...
the treatment target. Unlike photon beam arc therapy where the treatment isocentre is located at or near the centre of the target volume, the isocentre of electron arc therapy is located relatively far behind the target volume, i.e. a small focus to skin distance is used. The gantry rotates slowly during treatment delivery, and this delivers a narrow strip of uniform electron beam around the body surface beneath which the target volume is located. The gantry speed, gantry start and stop angles, dose per gantry angle rotation, isocentre location, and field length are chosen such that the treatment can deliver a uniform dose to the target volume. The treatment can have large penumbra. To better protect the critical normal tissues near the field edges, lead sheet of appropriate thickness and with appropriate cut-out is sometimes placed on the patient to improve the sharpness of the shielding. Another potential problem with the treatment is the possibility delivering a high x-ray dose given to the isocentre area. The x-ray dose is due to bremsstrahlung radiation that is focused at the isocentre during electron arc treatment.

**Electron backscatter factor**

*(Radiotherapy)* The presence of any inhomogeneity (e.g. lung, bone) within the path of a beam will significantly modify the electron dose distribution due to different absorption properties and effects on the electron scattering. Therefore, when an inhomogeneity of high atomic number (relative to water) is in the beam, electrons being backscattered from the inhomogeneity may result in an increased dose at the interface between the tissue and the inhomogeneity.

Examples where internal shielding is used to protect normal structures beyond the target include treatments of the lip, buccal mucosa or ear. Care must be taken as the electrons backscattered from the shielding (usually lead) can deliver a high dose to the healthy tissue in contact with the shielding. Therefore it is common to coat the shielding in a few millimetres of wax which will absorb the low energy scattered electrons.

To quantify the change in dose due to the backscattering a property known as the electron backscatter factor (EBF) may be determined. The EBF is defined as the ratio of the dose at the interface surface with and without the inhomogeneity present. The EBF has been found to increase with increasing atomic number and decrease with increasing beam energy.

**Abbreviation:** EBF = Electron backscatter factor.

**Related Articles:** Backscatter, Backscatter factor, Bone soft tissue interface, Inhomogeneity correction factor


**Electron beam**

*(Radiotherapy)* Linear accelerators are used to provide clinical beams of electrons which usually have nominal energies in the range 4–20 MeV. Electrons are produced by thermionic emission and injected into a waveguide structure operating in conjunction with microwave radiation at a frequency of 3000 MHz. The microwave energy required to accelerate the electrons is delivered to the accelerating structure in the form of short duration pulses at a repetition rate in the range 50–300 Hz. The electron beam, with a typical diameter in the range 4–6 mm, has a maximum energy and a narrow beam energy spread (a desirable feature), which depends on the accelerator design (full width half maximum 40 keV for a microtron and 50 keV for a linear accelerator). The beam then passes through the exit window and enters the beam handling and monitoring system of the accelerator. Linacs are equipped with a scattering and collimation system to achieve field flatness and symmetry for a 40 x 40 cm field at one meter. The most common scattering technique utilises a dual set of foils where the second foil is positioned at some distance from the first. The attractive feature of electron beams used in radiotherapy relates to the fact that the absorbed dose decreases sharply beyond a certain depth in the patient and this depth varies with beam energy. Electron beams can therefore be used to treat superficial cancer at depth up to about 5 cm and are generally as a single field irradiation.

**Electron beam CT**

*(Diagnostic Radiology)* The electron beam CT (EBCT) scanner is a novel concept of CT scanner design with no mechanical motion. It was originally marketed as the Imatron scanner, but currently is marketed by GE Medical Systems as the eSpeed.

In place of a rotating gantry with x-ray tube, the EBCT employs an electron beam which is accelerated and electromagnetically steered around an anode, consisting of four tungsten rings forming an arc of 210° around the patient (Figure E.21). One sweep of the arc results in a partial scan within a time of 33–100 ms. The x-rays produced at the target are collimated to a fan beam 47 cm diameter at the isocentre.

The detector array consists of two solid-state arcs of 216°, opposite the target rings. The scanner can therefore acquire two simultaneous slices when one target ring is used, or eight slices, if all four targets are used in sequence.

Its high temporal resolution made this scanner particularly suitable for cardiac applications. However, because of limitations in terms of power output, slice thickness and volume coverage, it has been largely superseded by the latest models of multislice CT scanners.

**Related Articles:** Imatron, Multislice CT scanner

**Electron beam dosimetry**

*(Radiotherapy)* The absorbed dose delivered by an electron beam can be measured by several methods but it is often determined by ionisation chamber measurements. The theoretical foundations of absorbed dose dosimetry in an electron beam are given by the Bragg–Gray cavity theory. Electron beam dosimetry is more complex than photon beam dosimetry because of the characteristics of the electron interaction with matter. Electrons are charged particles,
and, therefore, the electron energy spectrum varies with depth in a material because of their interaction. In addition the polarisation density effect becomes important for high energy electrons and consequently the dose conversion factor for an ionisation chamber in a medium varies with depth. Knowledge of the electron energy spectrum is crucial for absorbed dose determination since the fundamental physical parameters, such as the stopping power of a material, are a function of energy. Many international codes of practice and protocols have been published for the dosimetry of electron beams under reference conditions. It is essential, because of the dosimetric considerations, that a strict adherence to the used protocol is kept without mixing together recommendations from different protocols.

**Electron capture**
(Radiation Protection) Electron capture, sometimes known as K-capture, is where an electron is captured during radioactive decay of a radionuclide from one of the electron orbitals, most often the K-shell, of a radionuclide and combines with a proton in the nucleus with the formation of a neutron and a neutrino:

\[ P + e \rightarrow n + \nu \]

See article on Electron capture (EC) decay.

**Related Articles:** Electron capture (EC) decay, Neutron, Neutrino, Radioactive decay, Radionuclide

**Electron capture (EC) decay**
(Radiation Protection) Electron capture, sometimes known as K-capture, is a type of radioactive decay where an electron is captured from one of the electron orbitals, most often the K-shell, of a radionuclide and combines with a proton in the nucleus with the formation of a neutron and a neutrino:

\[ P + e \rightarrow n + \nu \]

The overall change in the nucleus is exactly the same as in positron decay. It is a different mechanism from positron decay (beta plus decay) in that no particle, apart from a neutrino, is emitted from the nucleus but the net effect in the nucleus is the same – one less proton and one more neutron.

An example is the decay of chromium-51 to vanadium-51. Ninety-one per cent of \(^{51}\text{Cr}\) atoms decay directly to the ground state of \(^{51}\text{V}\) and 9% via the 320keV level with resulting emission of a 320keV gamma ray. The K-shell electrons are involved in 90% of captures, the remainder from the L-shell (Figure E.22).

**Related Articles:** Beta decay, Electron, Neutron, Neutrino, Positron, Positron decay, Radioactive decay, Radionuclide

**Electron contamination**
(Radiotherapy) The photon beam exiting the Linear Accelerator Treatment head has some low energy electron contamination that arises from photon interactions (predominantly Compton) within the treatment head. These secondary electrons contribute to the dose received by the patient, particularly by increasing the surface dose. The range of contamination electrons for 21 MV photon beam is about 6cm. Therefore 10cm is recommended as reference depth in some dosimetric protocols.

Any scattering material placed into the beam increases the electron contamination and consequently reduces skin sparing. A noticeable effect of electron contamination is a reduction of build-up region occurring with increasing field size, due to the greater area of scattering material visible to the beam.

To reduce electron contamination, beam modifiers and accessory trays are generally placed as far away from the patient as possible.

**Related Articles:** Build-up region, Build-up dose

**Further Reading:** Dutreix, A., B. E. Bjargard, A. Bridier, B. Mijnheer, J. E. Shaw and H. Svensson. Monitor Unit Calculation for High Energy Photon Beams, Physics for clinical radiotherapy, Booklet No. 3. ESTRO (European Society for Therapeutic Radiology and Oncology), 1997, Brussels, Belgium.

**Electron density**
(Radiotherapy) The electron density is the number of electrons per volume unit. In Table E.2 electron densities for some tissues are reported. The treatment planning system needs the physical and geometrical characteristic of the patient to calculate the dose distribution with a specific radiation beam arrangement. The treatment planning systems utilise a 3D matrix which contains electron densities of every voxel within the area of interest to incorporate the patient heterogeneity into the dose calculation results. Information about electron densities is obtained from CT numbers, via the Hounsfield units. Published or empirically determined conversion tables are used for the conversion from CT numbers to electron densities. A formula is recommended as a default for use in the treatment planning systems in circumstances where no data are available for a particular scanner. To achieve the maximum accuracy, particular CT calibration curve for each CT scanner should be provided using phantom with inhomogeneities with known RED. In Figure E.23 a typical CT calibration curve is given. Usually calibration curves are bilinear.

**Abbreviations:** CT = Computed tomography, HU = Hounsfield unit and RED = Relative electron density.

<table>
<thead>
<tr>
<th>Material</th>
<th>Physical Density (g/cm(^3))</th>
<th>Electron Density per cm(^3) (\times 10^{23})</th>
<th>Electron Density Relative to Water (RED)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Water</td>
<td>1.00</td>
<td>3.340</td>
<td>1.000</td>
</tr>
<tr>
<td>Lung (inhale)</td>
<td>0.20</td>
<td>0.634</td>
<td>0.190</td>
</tr>
<tr>
<td>Lung (exhale)</td>
<td>0.50</td>
<td>1.632</td>
<td>0.489</td>
</tr>
<tr>
<td>Adipose</td>
<td>0.96</td>
<td>3.170</td>
<td>0.949</td>
</tr>
<tr>
<td>Breast (50/50)</td>
<td>0.99</td>
<td>3.261</td>
<td>0.976</td>
</tr>
<tr>
<td>Muscle</td>
<td>1.06</td>
<td>3.483</td>
<td>1.043</td>
</tr>
<tr>
<td>Liver</td>
<td>1.07</td>
<td>3.516</td>
<td>1.052</td>
</tr>
<tr>
<td>Trabecular bone</td>
<td>1.16</td>
<td>3.730</td>
<td>1.117</td>
</tr>
<tr>
<td>Dense bone (800 mg/cc)</td>
<td>1.61</td>
<td>5.052</td>
<td>1.512</td>
</tr>
<tr>
<td>Dense bone (1000 mg/cc)</td>
<td>1.66</td>
<td>5.243</td>
<td>1.570</td>
</tr>
<tr>
<td>Dense bone (1250 mg/cc)</td>
<td>1.83</td>
<td>5.718</td>
<td>1.712</td>
</tr>
<tr>
<td>Dense bone (1500 mg/cc)</td>
<td>2.00</td>
<td>6.209</td>
<td>1.859</td>
</tr>
<tr>
<td>Dense bone (1750 mg/cc)</td>
<td>2.17</td>
<td>6.698</td>
<td>2.005</td>
</tr>
<tr>
<td>Titanium</td>
<td>4.51</td>
<td>12.475</td>
<td>3.735</td>
</tr>
</tbody>
</table>

**FIGURE E.22** The decay of \(^{51}\text{Cr}\) by electron capture.
**Electron dual scattering foils** (Radiotherapy) The electron beam that exits the waveguide has a thin pencil beam shape, which would produce a highly localised dose without alteration. To create a clinically useful electron beam, two scattering foils of high atomic number are used within the treatment head of the linear accelerator, which firstly widen the beam, and secondly flatten it. The scattering foils are together with the flattening filters mounted on a rotating carousel or sliding drawer for ease of mechanical positioning into the beam, as required. The dual foil system improves the single foil system because equally uniform beams can be produced with a considerably smaller total foil thickness, which reduces the energy spread. However these scattering foils will contribute to unwanted Bremsstrahlung radiation, which must be accounted for by careful collimation of the beam (Figure E.24).

**Electron field effective width** (Radiotherapy) The effective width of an electron beam field is the distance between the points of 50% dose at the radiation beam axis along each of the major axes of the field. The distance is obtained by measuring the profiles of relative doses in a water phantom in a plane parallel to the surface of the water phantom and perpendicular to the central axis. The width of the electron beam is projected by the light field of the linear accelerator. In Figure E.25 isodose profiles of an electron beam are shown. In Figure E.25a, the profile is on a plane parallel to the incident beam direction containing the depth of maximum dose, the standard measurement depth (SMD). The base depth is on a plane parallel to the SMD plane containing the 90% point on the beam central axis. In Figure E.25b, the beam cross section is shown at two different depths: left half at SMD and right half at base depth.

**Electron fluence** (Radiotherapy) The electron fluence $\Phi$ is the quotient $dN$ by $da$, where $dN$ is the number of electrons that enter an imaginary sphere of cross sectional area $da$, i.e.

$$\Phi = \frac{dN}{da}$$

The fluence is usually expressed in units m$^{-2}$ or cm$^{-2}$.

**Electron gun** (General) In a cathode ray tube (CRT), an electron gun is used to produce a focused stream of electrons, which strike the screen phosphor (Figure E.26).

The components of the electron gun are the cathode and filament, control grid, focusing anode and accelerating anode. Electrons boil off the heated cathode filament through thermionic emission and...
are swept along the potential difference between the cathode and anodes to form a stream of electrons. The control grid voltage is used to control the beam intensity: when the grid is positive with respect to the filament, electrons are accelerated toward the screen; when the grid is negative with respect to the filament, no electron current is flowing toward the screen. The accelerating anode is at a positive potential relative to the cathode of several thousand volts. The focusing anode ensures the electron beam is brought to a focus on the CRT screen.

In a colour CRT, a separate electron gun is allocated to produce red, green and blue colours on the screen. A mask is used to ensure only electrons from the correct gun excite phosphor dots of the corresponding colour on the CRT screen.

**Electron hole pair**  
*(General)* In an energy level description of conduction, the *conduction band* represents the electron energy levels where conduction can occur. The *valence band* for a material represents the upper electron energy states occupied at absolute zero. In a metal, the conduction and valence bands overlap, and conduction can take place freely. In a semiconductor, there is an energy gap between the valence and conduction bands. All possible electron states are occupied at the top of the valence band in a semiconductor. Conduction cannot take place within the valence band as all states are occupied (Figure E.27).

For conduction to take place in a semiconductor, an electron in the valence band must absorb sufficient energy to be promoted to the conduction band. Promotion of an electron to the conduction band leaves behind an unoccupied state in the valence band called a ‘hole’. As other electrons in the valence band can now move to occupy this state, in turn leaving behind a hole, holes can be thought of as charge carriers in a semiconductor. Promotion of an electron to the conduction band by, e.g., absorption of radiation generates an ‘electron hole pair’. Electron hole recombination occurs where a free electron falls from the conduction band and fills a hole in the valence band (Figure E.28).

**Electron maximum range**  
*(Radiotherapy)* The maximum range ($R_{\text{max}}$) is defined as the maximum depth to which electrons can penetrate the material they are incident upon. It is defined as the point at which the central axis depth dose curve meets the Bremsstrahlung x-ray contamination (see Figure E.29). This is not necessarily a well-defined measurement point, and the practical range tends to be used more commonly.

*Abbreviation:* PDD = Percentage depth dose  
*Related Articles:* Electron ranges, Electron practical range, Electron therapeutic range

**Electron oblique incidence**  
*(Radiotherapy)* Typically radiotherapy beam data (e.g. depth dose etc.) are measured with the beam incident perpendicular to the flat and uniform surface of the phantom. However, this situation occurs rarely. Hence these data cannot be used for dose distribution calculations directly.

For small angles (of typically less than 20°) the isodose lines tend to simply follow the angle of the surface contour. For angles greater than 20° there are increasingly significant changes to the depth dose characteristics – the depth of maximum dose decreases with increasing angle, the dose at shallow depths increases, and the dose at greater depths is reduced relative to the standard normal incidence data. At large angles of incidence the dose at the maximum depth increases significantly.
The most common method for understanding this problem is to use a pencil beam algorithm for dose calculations.

**Related Article:** Oblique incidence


**Electron off axis factor**

(Radiotherapy) The off axis factor (off axis ratio – OAR) represents the ratio of the dose at a point a certain distance from the beam central axis and the dose at a point on the central axis in the plane perpendicular to the beam axis where both points are placed. Off axis factors can be measured at any depth and a plot of off axis factors against distance from central axis gives the beam profile at that depth. Thus off axis factors follow changes in the beam flatness as a function of the depth and the distance from the central axis.

**Abbreviation:** OAR = Off axis ratio.

**Related Articles:** Beam flatness, Beam symmetry, Off axis dose distribution

**Electron positron pair**

(Radiation Protection) See Pair production

**Electron practical range**

(Radiotherapy) The electron practical range ($R_p$) is found by using a plot of the central axis percentage depth dose curve (see Figure E.30). First draw a tangent to the depth dose curve at the depth of 50% dose. Next extend the Bremsstrahlung contamination tail back parallel to the depth axis towards the dose axis. The point at which these two lines meet is the electron practical range (depth). The practical range increases with increasing beam energy and can be approximately found by using the following assumption:

Practical range (in mm) is equal to five times the beam energy (in MeV), i.e. for a 6 MeV beam the practical range should be approximately 30 mm.

Conversely the practical range (in cm) can also be used to calculate the most probable beam energy at the surface ($E_{p,0}$) by the following relationship:

$$E_{p,0} \text{ (MeV)} = 0.22 + 1.98R_p + 0.0025R_p^2$$

It is also possible to calculate the mean electron energy (MeV) at depth ($E_i$) from the practical range (cm) and the mean electron energy (MeV) at the surface ($E_o$) using the following equation:

$$E_i = E_o \left(1 - \frac{z}{R_p}\right)$$

**Related Articles:** Electron ranges, Electron maximum range, Electron therapeutic range

**Electron ranges**

(Radiotherapy) It is possible to characterise electron beams used in radiotherapy by the depth at which a certain point on the depth dose curve is located. Typical values are depth of maximum dose, 90% dose, 80% dose, 50% dose and the depth at which the dose is reduced to the value of the proportion of contamination x-rays. There is a series of simple approximations which relate each of these depths to the nominal energy of the electron beam, and these are listed as follows:

- Depth of maximum dose (mm) is approximately two times the beam energy (MeV).
- Depth of 90% dose (Therapeutic range) (mm) is approximately three times the beam energy (MeV).
- Depth of 80% dose (cm) is approximately one-third of the beam energy (MeV).
- Depth of 50% dose (mm) is approximately four times the beam energy (MeV).
- Electron practical range (mm) is approximately five times the beam energy (MeV).

Therefore applying these approximations for the example of a 6 MeV electron beam will give the following results:

The depth of maximum dose is at 12 mm, the 90% dose is at 18 mm, the 80% dose is at 20 mm, the 50% dose is at 24 mm, and the practical range is at 30 mm.

Knowledge of such depths is used when selecting the appropriate energy for a treatment. Other equations can be used, e.g. to relate the practical range to the most probable beam energy at the surface or to the mean electron energy at depth (see the article Electron practical range for detail on these equations).

**Related Articles:** Electron maximum range, Electron practical range, Electron therapeutic range

**Electron spin**

(General) Experiments by Otto Stern and Walter Gerlach in 1921 have shown that the electron, in addition to its orbital angular momentum $\vec{L}$, possesses an intrinsic angular momentum. This intrinsic angular momentum is referred to as the spin $\vec{S}$ and is specified by two quantum numbers: $s = \frac{1}{2}$ and $m$, that can take two values (1/2 or −1/2).

The electron spin is given as

$$S = \hbar \sqrt{s(s+1)} = \frac{\sqrt{3}}{2} \quad \text{and} \quad \text{its } z \text{ component } S_z = m\hbar$$

The spin quantum number is one of four quantum descriptors of the atom: principal, orbital, magnetic and spin quantum numbers. The spin quantum number $s$ for an electron is $\frac{1}{2}$, its $z$ component $m$ can take two values $+\frac{1}{2}$ and $-\frac{1}{2}$.

Electron spin explains hyperfine lines in the hydrogen spectrum and the outcome of the Stern–Gerlach experiment.

**FIGURE E.30** An illustration of the definition of the electron practical range.
In the classical conception of spin, the electron is viewed as a ball of charge spinning on its own axis. While this may be a useful visual aid, spin is a quantum phenomenon and experimental observations are not consistent with this simple mechanical view.

**Electron stopping power**

(Radiotherapy) The stopping power for electrons and positrons is different from that of heavy charged particles because an electron loses a large fraction of its energy in a single collision with an atomic electron which has equal mass and also because the electron is identical to the atomic electron with which it collides. Quantum mechanics implies that incident and struck electron cannot be distinguished. The collisional stopping power for electrons and positrons can be written as follows:

\[
\left( \frac{dE}{dx} \right)_{\text{coll}} = \frac{4\pi\kappa_0 e^4 n}{mc^2} \left[ \ln \left( \frac{mc^2 \sqrt{1 + \beta^2}}{\sqrt{2I}} \right) + F^\pm(\beta) \right]
\]

where

\[
F^\pm(\beta) = \frac{1 - \beta^2}{2} \left[ 1 + \frac{\tau^2}{8} - (2\tau + 1)\ln 2 \right]
\]

for electrons and

\[
F^\pm(\beta) = \ln 2 - \frac{\beta^2}{24} \left[ 23 \frac{14}{\tau + 2} + \frac{10}{(\tau + 2)^2} + \frac{4}{(\tau + 2)^3} \right]
\]

for positrons.

The formula is obtained by combining soft collisions and hard collisions using the Møller’s electron cross section and Bhabha’s positron cross sections for free electrons.

The total stopping power for electrons and positrons is the sum of the collisional and radiative stopping powers:

\[
\left( \frac{dE}{dx} \right)^\pm = \left( \frac{dE}{dx} \right)_{\text{coll}} + \left( \frac{dE}{dx} \right)_{\text{rad}}
\]

The following approximate formula gives the ratio of radiative and collisional stopping powers for an electron of total energy \(T\), expressed in MeV, in an element of atomic number \(Z\):

\[
\frac{(-dE/dx)_{\text{rad}}}{(-dE/dx)_{\text{coll}}} \approx \frac{ZT}{800}
\]

**Electron therapeutic range**

(Radiotherapy) The therapeutic range \(R_{90}\) is defined as the electron isodose that covers the distal boundary of the tumour to be treated. Typically the electron energy to be used in a treatment is chosen such that the 90% (or 85% depending on local practice) dose level beyond the peak is the isodose that reaches this distal boundary. The therapeutic range increases with increasing energy. A simple approximation for remembering the depth of 90% dose is that the depth of 90% dose (in mm) is equal to three times the beam energy (in MeV), i.e. for a 6 MeV beam the depth of 90% dose is 18 mm (Figure E.31).

**Related Articles:** Electron ranges, Electron maximum range, Electron practical range

**Electron transport**

(Radiotherapy) Much of the energy transferred to the medium in Radiotherapy is through the electrons when an incident photon (causing an ionisation event) releases an electron and the latter in turn releases secondary electrons. The electrons contribute to energy deposition along the paths taken. At photon beam energies below 6 MeV electron equilibrium is assumed to hold (AAPM, 2004). This means that the number and energy distribution of electrons entering a local volume around the primary photon interaction are equal. In addition the dose contribution of scattered photons dominates over electrons. Above 6 MeV these assumptions no longer hold and the dose profile deposition is altered due to electron transport. In dosimetry, mathematical models of energy deposition due to electron transport have been developed (Figure E.32).


**Electron volt**

(General) The electron volt (eV) is a unit of energy. It is the standard unit to describe x-ray and gamma ray energies in diagnostic.
radiology and radiotherapy as well as energies of atomic and subatomic particles.

One electron volt is the energy gained when an electron is swept across a potential difference of 1 V. It is a convenient unit for use as the definition of the electron volt can be directly related to the basic physics of an x-ray tube or a linear accelerator. For example, if 70kV is applied across an x-ray tube, electrons are swept across the tube and strike the anode with a maximum energy of 70 keV. The maximum x-ray energy in keV is then numerically equivalent to the maximum voltage across the tube in kV:

\[ 1 \text{eV} = 1.602 \times 10^{-19} \text{J} \]

Electronic binding energy

*(General)* The binding energy of an electron in an atom is the energy required to remove the electron from the atom. The binding energy is characteristic of the atom and the 'shell' from which the electron is removed. Electrons in the orbits closer to the nucleus are more tightly bound and have higher binding energies. For example, the K and L shells of tungsten have binding energies of 69.5 and 12.1 keV, respectively. In atoms, binding energy ranges from a few electron volts required to ionise an alkali atom by removing its outer shell electron to about 150 keV required to remove a K shell electron from a very high atomic number atom.

Electronic equilibrium

*(Radiotherapy)* The absorbed dose deposition by photons through interaction with matter is a two step process in which they first transfer their energy to electrons (charged particles) which in turn deposit energy in the matter. However, the energy loss by photons at a particular location is not necessarily equal to the absorbed dose by the medium at that location. This is because the secondary electrons (charged particles) travel a certain distance from the interaction point before they release all their kinetic energy. Electronic equilibrium exists for a volume \( v \) if each electron of a given energy leaving \( v \) is replaced by an electron of the same energy entering the volume (Figure E.33).

In the volume where the electronic equilibrium exists

\[ D = K_r = \Psi \left( \frac{\mu_{en}}{\rho} \right) \]

where

- \( D \) is the dose
- \( K_r \) is the collision kerma
- \( \Psi \) is the energy fluence
- \( \mu_{en}/\rho \) is the mass energy absorption coefficient

If the photon fluence changes over a distance comparable with the range of the secondary electrons in the matter because of attenuation or if the volume considered is near an interface between different media, i.e. within the range of secondary electrons from the interface, then the dose at the volume considered is due part to electrons originating in a region of different photon fluence or of different composition. In this case the dose in the volume element will not be equal to the energy removed from the radiation beam in that volume.

In Figure E.34a one hundred secondary electrons with a range \( R \) are supposed to be set in motion in any square from A to G, without any photon beam attenuation. The dose increases with depth to reach a plateau while the photon fluence is constant. In this situation the electronic equilibrium can be obtained in any point beyond the build-up region. In Figure E.34b a fluence attenuation of 5% will reduce by the same amount the secondary electrons in any square from A to G. The electronic equilibrium cannot be obtained because of the photon attenuation and the subsequent variation of \( \Psi \). For depth beyond the build-up region, the \( D \)-curve becomes parallel to the \( K_r \)-curve and a condition called Transient electronic equilibrium is obtained where

\[ D \cong K_r (1 + \mu' \tau) \]

**FIGURE E.33** Electronic equilibrium.

**FIGURE E.34** Illustrating (a) electronic equilibrium condition and (b) transient electronic equilibrium.
where
\[ \mu' \] is the common slope of the D- and K-curves
\[ x \] is the mean distance the secondary electrons carry their kinetic energy in the direction of the photons

Electronic generators
(Nuclear Medicine) Electronic generators are automated synthesis systems which are integrated with compact cyclotron technology for the preparation of radiopharmaceuticals. These are for use in preclinical and clinical investigations with positron emission tomography. The system is usually controlled by a PC.

Electronic medical record (EMR)
(Diagnostic Radiology) The electronic medical record (EMR) is a digital version of the medical record – a systematic collection of data related to patient’s medical history and care.

Most common types of data stored under the EMR:
- Patient demographics
- Medical history
- Examination data
- Progress reports
- Allergy lists
- Immunisation status
- Laboratory test results
- Radiology images (x-rays, CT, MRI, US, etc.)
- Other clinical images – endoscopy, laparoscopy, clinical photographs
- Medication information
- Evidence-based recommendations for specific medical conditions
- Administrative data – record of appointments, reminders, billing records

EMRs are handled using distributed information systems (IS) within the enterprise (such as hospital information system [HIS], picture archiving and communication system [PACS], etc.). The application and scope is dependent on the particular IS, application, enterprise structure, functional and operational levels, etc.

Electronic portal imaging
(Radiotherapy) In external beam radiotherapy, portal imaging involves using the radiation from the treatment beam (i.e. the treatment portal) that passes through the patient to form an image. The variations in absorption across the field, particularly between bone, soft tissue and the field edge produce the image.

Film versus Electronic Detectors: Traditionally film has been used. More recently electronic detectors have been developed and now largely replace film. These have the advantage of ease of use (e.g. no need to process the image), the information is available immediately and may be analysed automatically.

Information Measured: Electronic portal imaging (EPI) is used to monitor the accuracy of treatment delivery, usually by determining the positioning of bony anatomy relative to the edge of the radiation field. This may be compared with a predicted image from a treatment simulator or a DRR generated from the TPS. The clinical target volume may move relative to the bony structure and hence the use of bony structures for set-up may result in uncertainty in the positioning of the target volume. Figure E.35a shows a DRR for a prostate treatment and Figure E.35b the corresponding electronic portal image.

Comparison with Diagnostic Imaging: Images for EPI are generally of lower quality than diagnostic, kV energy images because of the high energy used of several MV

Abbreviations: DRR = Digitally reconstructed radiograph, EPI = Electronic portal imaging, kV = Kilovoltage, MV = Megavoltage and TPS = Treatment planning system.

Related Articles: Electronic portal imaging device, Port film, Online portal imaging, Digitally reconstructed radiograph (DRR)


Electronic portal imaging device
(Radiotherapy) An electronic portal imaging device (EPID) is a detector that produces a 2D image of the treatment portal, usually with the patient present in the field. The images obtained are generally of megavoltage energy and thus of lower quality than diagnostic, kilovoltage energy images. Several technologies have been developed for EPIDs: camera system; liquid ionisation chamber arrays and flat panel detectors

Camera Systems: Some of the first systems to be developed were based on the use of cameras. Figure E.36 shows a schematic diagram of such a detector. X-rays transmitted through the patient interact with a fluorescent screen to produce light. The mirror takes
Electroscope

An electroscope is an early scientific instrument for detecting the presence and magnitude of electric charge or ionizing radiation. The device's operation is based on the Coulomb force law and detects the electric charge by the motion of a test object due to the Coulomb electrostatic force. There are various types of electrosopes. The most common type is the gold-leaf electroscope, in which two thin gold leaves are suspended from a conducting rod held in an insulated container. When a source of static electricity is brought near to the rod, some of the electrons in the rod are pushed to the leaves if the source is negative or pulled up to the rod from the leaves if the source is positive. In both cases the leaves are charged likewise and so they repel each other and

Related Articles: Electronic portal imaging device, Port film, Online portal imaging, Digitally reconstructed radiograph (DRR)

Electrostatic deflector

(Nuclear Medicine) The electrostatic deflector is used in a positive particle cyclotron to redirect the particle beam (typically protons) onto a target. In a positive particle cyclotron, the electrostatic deflector has a negative potential in order to extract the beam from the magnetic field.

Related Articles: Cyclotron, External beam irradiation

Elementary particles

(General) The term elementary particle refers to a subatomic particle with a definite mass and charge which contains no discernible substructure. Many families and sub-families of elementary particles exist. Three basics quantities are used to identify the particles: mass, charge and spin. According to their masses and the dominant interaction particles are grouped into four families: bosons (graviton and photon, massless bosons), leptons (fermions, light particles), mesons (bosons, intermediate mass), baryons (fermions, heavy particles). Fermions, having half-integer spin, obey to the exclusion principle and bosons do not because they have integer spin. Baryons and mesons are subject to all four interactions: strong, electromagnetic, weak and gravitational. Leptons are not sensitive to strong interaction while photons (massless bosons) are sensitive to electromagnetic interaction.

Embryo

(General) In humans the term embryo is used to describe the developing baby from the moment of fertilisation until the end of the 8th week, after this the term used is foetus.

Related Article: Foetus

Emission computed tomography (ECT)

(Nuclear Medicine) Tomography is a method to reconstruct transversal image from projections of data acquired in different angles around the patients. The reconstruction is usually made by filtered backprojection or iterative methods. In nuclear medicine measurements are made with either radionuclides that emit single photons or with positron emitters that produce two 511 keV photons from positron annihilation. These two imaging modalities are referred to as single-photon emission computed tomography (SPECT) and positron emission tomography (PET). The family name for these modalities is emission computed tomography (ECT).

Emulsion layer

(Diagnostic Radiology) See Film emulsion

End diastolic velocity

(Ultrasound) Diastole describes the relaxation of the chambers of the heart. The end diastolic velocity is the velocity in the systemic arterial circulation at the end of ventricular diastole immediately preceding the onset of systole. In the case of a flow in an artery leading to high terminal resistance, this value may be zero (Figure E.40).

Abbreviations: EDV = End diastolic velocity and PSV = Peak systolic velocity.

Related Article: Peak systolic velocity

End of bombardment (EOB)

(Nuclear Medicine) End of bombardment (EOB) refers to the moment when irradiation stops, i.e. when the beam is turned off on particle accelerators such as cyclotrons.

Related Article: Avogadro’s number


Energy absorption coefficient

(Radiation Protection) The energy absorption coefficient, \( \mu_a \), is the fraction of the energy of a gamma or x-ray beam that is absorbed per unit distance in the medium. Since a fraction of the incident beam is scattered, the absorbed energy is only a part of the total energy lost by the beam:

\[
I = I_0 \exp[-(\mu_a + \mu_s)t]
\]

where

- \( I_0 \) is the incident radiation intensity at the surface of the medium
- \( I \) is the radiation intensity transmitted through a medium of thickness \( t \)
- \( \mu_s \) is the fraction of the beam scattered per unit distance

The energy absorption coefficient, as defined earlier, is often called linear energy absorption coefficient and has dimension L\(^{-1}\) (usually given in cm\(^{-1}\)). It depends on the radiation energy, on the composition and on the density of the medium.
Energy deposition

(Radiation Protection) At a cellular level, damage is caused to the human body by the deposition and absorption of the energy inherent in an incident radiation beam, whether it be the energy of ionising or non-ionising photons, or the kinetic energy of the atomic particles (alpha, beta, protons, neutrons, etc.) in particle radiation.

The nature and severity of damage caused to individual cells, and to the tissue or organ made up of those cells, can be better understood by modelling the concentration and distribution of energy deposition within the cells. Such modelling helps to definition any radiation as having either a high linear or low linear transfer characteristic, with associated radiation weighting factor.

As the deposition of energy in matter has statistical fluctuations, energy deposition is stochastic. Thus energy deposition within a mass of material may be estimated by averaging observed values of the stochastic variations.

The energy deposited in a single interaction, $i$, can be defined at the point of interaction:

$$
\varepsilon_i = \varepsilon_{in} - \varepsilon_{out} + Q
$$

where

$\varepsilon_{in}$ is the energy of the incident ionising particle (excluding rest energy) (expressed in Joules or electron-volts (eV))

$\varepsilon_{out}$ is the sum of the energies of all particles leaving the interaction (excluding rest energy)

$Q$ is the change in the rest energies of the nucleus and of all particles involved in the interaction

$Q$ can be either +ve or −ve, indicating either a decrease or an increase, respectively, in the net rest energy; of all the particles resultant from the interaction

Related Articles: Absorbed dose, Linear energy transfer, Radiation weighting factor

Energy fluence

(Radiation Protection) Energy fluence, usually denoted by the Greek letter $\psi$, is one of the units used to describe a radiation field. It is the radiance (product of number of particles, $N$, and particle energy, $E$) per unit area.

Hence

$$
\psi = \frac{dR}{da}
$$

The unit of energy fluence is J/m$^2$.

Related Article: Incident energy fluence

Energy gap

(Nuclear Medicine) See Band gap

Energy loss rate

(Radiation Protection, General) The rate of loss of energy from a beam of ionising radiation as it traverses an absorbing medium is described by the linear energy transfer of the type of radiation and the absorption characteristics of the medium.

For more details, see articles on Absorption, Attenuation and Linear energy transfer.

Related Articles: Absorption, Attenuation, Linear energy transfer

Energy resolution

(Radiation Protection) Energy resolution is the ability of the radiation detector of a measuring system (e.g. spectrometer) to discriminate photons and particles with different energies.

The energy resolution (in %) is defined by the full width at half maximum (FWHM) of a pulse (photopeak) divided by the energy at the maximum of the pulse (Figure E.41):

$$
\text{Energy resolution}(%) = 100 \times \frac{\text{FWHM}}{E_{\text{max}}}
$$

Semiconductor detectors (e.g. Germanium detector) have very narrow pulses, and their Energy resolution is very good (below 1%). This makes them very suitable for spectrum analysers. Scintillation detectors have Energy resolution of the order of 8%–9%.

Abbreviation: FWHM = Full width at half maximum.

Related Article: Multichannel analyser

Energy spectrum

(Diagnostic Radiology) The energy spectrum is the range and distribution of photon energies in an x-ray beam.

See also Beam spectrum.

Related Article: Beam spectrum

Enhancing filter

(General) An enhancing filter is any device which alters the data being passed through in such a way as to enhance or make more prominent the information required, whilst reducing or limiting other data.

Enhancing filters may be made of a physical layer of material through which the signal passes (e.g. optical or x-ray) on its way to the detector. Optical filters may enhance specific colours or limit resolution, whilst x-ray enhancing filters may be made out of appropriate metals to harden or soften x-ray beams, to better image specific tissues.

Enhancing filters may also be used to post-process digital images. Many common image enhancement software packages exist for
Ensemble length
(Ultrasound) The term ensemble length is used to describe the number of pulses used to produce each colour ‘line’ in a colour flow image. Colour flow systems typically use autocorrelation methods to compare the shift in phase between pulses and thereby the Doppler frequency. SNR and therefore the quality and accuracy of the colour flow output generally improve with increased ensemble length. However, increasing ensemble length leads to time constraints with a possible reduction in frame rate.

In some systems, ensemble length may be altered by the operator to improve the sensitivity of the colour flow image at the expense of frame rate. The converse is also possible. The nomenclature of this control varies from scanner to scanner. Ensemble length is one of the many parameters, including line density, frame rate colour box width and depth and pulse repetition frequency which are varied automatically to maintain acceptable frame rate if changes are made to one or more of the others by an operator.


Entrance dose
(Radiotherapy) Entrance dose is the quantity which is often measured in the process of in vivo dosimetry. Within in vivo dosimetry, doses derived from the signal of the detector placed on the skin are compared with the theoretical values calculated by the treatment planning system (TPS). As the accuracy of the skin dose calculation is questionable, and in many cases irrelevant, the signal of the detector is related to the dose at a point which is still close to the skin, but at a certain depth where the accuracy of the TPS is much more satisfactory. At the entrance side of a medium irradiated by a single beam, the dose gradually increases from a low value at the surface up to a maximum value \( D_{\text{entrance}} \) at a depth \( d_{\text{max}} \) which depends upon the energy, the collimator opening, the skin-source-distance, the introduction of beam modifying devices and the distance separating them from the patient skin, etc. The measurement of \( D_{\text{entrance}} \) must be carried out with enough material in front and around the detector placed at skin level in order to establish the electron equilibrium.

Abbreviation: TPS = Treatment planning system.

Related Articles: Exit dose, Build-up, Build-up region, Diode detectors, In vivo dosimetry, Charged particle disequilibrium


Entrance surface air kerma (ESAK)
(Radiation Protection) The air KERMA (Gy) measured at the surface of the body. ESAK is useful in relation to estimating entrance surface doses, and hence skin doses, to patients.

A typical experimental set-up might measure ESAK with an ionisation chamber at the position where the surface of the body of the patient would be in an x-ray exposure, from which and entrance surface dose (Gy) and a skin dose (Sv) could be calculated using Monte Carlo–based conversion factors to take into account the backscatter from the underlying tissues of the body.

Related Articles: Air kerma, Entrance surface dose, Skin dose

Entrance window
(Diagnostic Radiology) See Image intensifier

EPI (echo planar imaging)
(Magnetic Resonance) See Echo planar imaging (EPI)

EPISTAR
See Echo-planar imaging and signal targeting with alternating radiofrequency

Equalisation
(Diagnostic Radiology) Equalisation is a general term for modifying signals or images to provide a corrected or more equal representation of the original data.

In time-varying signals such as sound recordings, equalisation refers to the adjustment (in frequency and phase) of the signals to correct for the distortion in their recording, transmission or storage.

In images, equalisation may be used to correct for imperfect ‘illumination’ or to optimise illumination and contrast such as through the use of ‘histogram equalisation’.

In certain x-ray imagers (e.g. mammography) ‘Scanning Equalisation’ may be used, which delivers a non-uniform distribution of x-ray exposure so as to maintain optimal sensor brightness and contrast over the area being imaged.

Equilibrium absorbed dose constant
(Nuclear Medicine) This parameter states the amount of radiation energy emitted from the radionuclide in the source organ. The energy emitted per unit of cumulated activity is called the equilibrium absorbed dose constant \( \Delta \). This factor needs to be calculated for each radiation type and it is given by

\[
\Delta = 1.6 \times 10^{-13} N E_i \text{ Gy} \cdot \text{kg/Bq} \cdot \text{s(J)}
\]

where

\( E_i \) is the average energy (MeV) of the \( i \)th emission

\( N_i \) is the number of particles or photons emitted per disintegration

The product between the cumulated activity and the equilibrium absorbed dose constant is the total energy radiated by the \( i \)th emission while activity is present in the source organ.

Another step to determine the absorbed dose to a target organ is to calculate the absorbed fraction (see separate article). The absorbed fraction and the equilibrium absorbed dose constant are often combined into a mean dose per cumulated activity \( S \) to simplify the calculation procedure. \( S \) is determined by the emission type, radiation energy and anatomic relationship and is determined for each source-target pair and radionuclide.

Related Articles: MIRD formalism, Absorbed fraction, Cumulated activity, Mean dose per cumulated activity


Equilibrium dose distribution
(Radiation Protection) This can be best understood in reference to Radiotherapy. The dose that is prescribed to the target area will be equal over that particular area and then decrease around the area, it is important that the distribution of the dose over the target area is equal to achieve optimal results, and this is known as equilibrium dose distribution. This can be achieved by using the multi-leaf/step collimators and wedges.
Equivalent dose ($H_T$)

(Radiation Protection) Equivalent dose is a unit used in radiation protection. It is the product of the absorbed dose $D$ (in Grays) to an organ or tissue and a factor, $W_e$ called the radiation weighting factor, a dimensionless quantity that characterises that damage associated with the relative biological effectiveness of different types of radiation.

$$H_T = D \times W_e$$

Equivalent dose has the unit of the Sievert (Sv).

The radiation weighting factor reflects the relative probability of damage to DNA arising from the same absorbed dose from different types of radiation due to the density of ionisation events along the track of the initiating photon or particle and subsequent energy transfer to the tissue. The following table gives the current weighting factors:

<table>
<thead>
<tr>
<th>Type of Radiation</th>
<th>$w_e$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Photons</td>
<td>1</td>
</tr>
<tr>
<td>Electrons, betas and muons</td>
<td>1</td>
</tr>
<tr>
<td>Protons</td>
<td>2</td>
</tr>
<tr>
<td>Alpha particles and heavy nuclei</td>
<td>20</td>
</tr>
<tr>
<td>Incident neutron and heavy nuclei</td>
<td>Continuous function – See ICRP Report 92</td>
</tr>
</tbody>
</table>

Therefore 1 Gy absorbed dose to a tissue from x-rays or beta particles gives rise to 1 Sv equivalent dose to the tissue.

Regulatory dose limits set for individual organs or tissues are given in terms of equivalent doses and are designed to ensure that the equivalent doses received by the organs or tissues are below the thresholds for deterministic effects.

**Related Articles:** Absorbed dose, Radiation weighting factor, Deterministic effects, Relative biological effectiveness


Equivalent field size

(Radiotherapy) Equivalent field size is a concept used to determine the size of the square radiation beam that has the same radiation dose output and percentage depth dose characteristic as a non-square radiation beam.

For further information see Equivalent square.

**Related Article:** Equivalent square

Equivalent mass of radium

(Radiotherapy, Brachytherapy) Calibration of source strength is a very important part of a comprehensive brachytherapy quality system. The instruments, ion-chambers and electrometers, used for source strength determinations, should have calibrations that are traceable to national and international standards.

**Specification of Source Strength for Photon Emitting Sources:** Source strength for a photon emitting source can be given as a quantity describing the radioactivity contained in the source or as a quantity describing the output of the source:

1. Specification of contained activity
   a. Mass of radium; mg Ra
   b. Contained activity; Ci, Bq
2. Specification of output
   a. Equivalent mass of radium; mg Ra eq
   b. Apparent activity

When artificial radionuclides became available, the brachytherapy community aimed at specifying source strength for these ‘radium substitutes’ in a radium-like manner. The new sources were similar in shape and strength to the old ones, and thus all the experience gained from the earlier radium treatments could easily be transferred.

The equivalent mass of radium, the mgRaEq, for an encapsulated photon emitting source is the mass of Ra-226 filtered by 0.5 mm Pt that gives the same air kerma rate or exposure rate at the same distance from the centre of the source. (Note that this could lead to interpretation problems for Ra sources. Consider a Ra tube with strength 20 mg and filtered by 1 mm Pt, not 0.5 mm Pt; the strength of the tube will correspond to 18.7 mgRaEq.)

In modern brachytherapy dosimetry, reference air kerma rate or air kerma strength is the quantity used to calculate absorbed dose. See Source strength for a full description of specification of source strength.

**Related Articles:** Source strength, Mass of radium, Contained activity, Apparent activity, Reference air kerma rate (RAKR), Air kerma strength

Equivalent square

(Radiotherapy) Depth dose data is generally only given for square field sizes, as it is impractical to measure every single possible field size and shape. To obtain values for rectangular or irregular fields, the field shape in question needs to be correlated with an equivalent square (or an equivalent field), a shape that has the same scatter/primary ratio (SAR) at the point in question.

During the commissioning of a linear accelerator, depth dose data for rectangular fields should be measured twice using different jaw pairs to define the field. This is because the machine output may be influenced by the upper or lower jaw position – the Collimator exchange effect.

**Circular Fields:** Circular fields have the same SAR as a square field of the same area, which roughly equates to an equivalent square of 90% of the diameter of the circular field.

**Rectangular Fields:** Rectangular fields generally give smaller depth doses than a square field of the same area due to the greater attenuation of the scatter from the extended sides of the rectangle. Hence to obtain the equivalent square size, the ratio of area to perimeter is normally equated. This formalism is known as Sterling’s approximation, and is described by the formula in Equation E.3. This approximation becomes less reliable for large fields $(a,b > 20 \text{ cm})$, and more accurate values can be found in the BJR supplement:

$$c = \frac{2ab}{(a+b)}$$

(E.3)

Equation E.3 is Sterling’s approximation for the equivalent square of side $c$, for a rectangle of sides $a$ and $b$. 
**Irregular Fields:** For irregular fields, the equivalent square can be determined by separating the blocked areas into composite rectangular fields, for which the equivalent square is known. Each composite equivalent square is then squared and summed, to form the ‘equivalent blocked area’. The square root of the difference between the equivalent blocked area and the open field area is the equivalent square of the blocked field.

For more complicated fields, Clarkson’s scatter integration method can be used, which is the most accurate, although the most labour intensive. This divides any irregular field into $N$ equal narrow sectors of circles with varying radii. The SAR is calculated as the average SAR of all $N$ circles, and associated equivalent square size found.


**Equivalent tissue air ratio (ETAR)**

*(Radiotherapy)* The Equivalent tissue air ratio (ETAR) method is a sophisticated 3D technique used to correct isodose data for tissue heterogeneity corrections for use in treatment planning.

It uses electron density information from 3D CT scans to calculate a weighted average of the electron densities, which is used to predict the scattering effect of the surrounding structures.

Other methods for tissue heterogeneity corrections include the ‘effective depth’ method, and the ‘Batho’ or ‘power law’ method. These are summarised in Report 24 from the International Commission on Radiation Units and Measurements.

**Abbreviations:** CT = Computed tomography and ETAR = Equivalent tissue air ratio.

**Related Articles:** Heterogeneity, Treatment planning system


**Erect**

*(General)* There are a series of terms used to describe the position of an individual when undertaking different imaging examination. Erect: Standing or sitting up. For example, an erect chest x-ray.

**Related Article:** Patient position

**Ernst angle**

*(Magnetic Resonance)* The Ernst angle is the flip angle which maximises signal in a spoiled gradient echo sequence for a given TR time and a given tissue T1. The Ernst angle is given by

$$\theta_{\text{ernst}} = \cos^{-1}(e^{-\text{TR}\cdot T1})$$

An optimum flip angle exists because two competing factors are at work in the relationship between signal amplitude and flip angle in a spoiled gradient echo sequence. A smaller flip angle causes longitudinal magnetisation to settle into a stronger steady state. A stronger steady-state longitudinal magnetisation favours a stronger signal as more magnetisation is available to flip onto the transverse plane. However, against this, a smaller flip angle flips less of the longitudinal magnetisation onto the transverse plane in each TR period, making less available for signal formation. The Ernst angle (named after R. Ernst) is the balance point between these competing factors, where signal is maximised.

**Escape peak**

*(Nuclear Medicine)* This is a peak in the energy spectrum of a NaI (Tl). Following an interaction with an iodine atom in a NaI (Tl) crystal, a characteristic iodine K-x-ray ($E = 30\text{keV}$) is emitted. For example, a photoelectric absorption of a $^{197}\text{Hg}$ gamma ray ($E = 77.3\text{keV}$) followed by an escape from the detector by the characteristic K-x-ray results in a registered energy of 47.3 keV. The effect is most prominent when acquiring low energy gamma rays with a thin crystal because the K-x-ray is more likely to escape. Also the relative distance between the two peaks gets smaller when the gamma ray energy increases.


**Escargot curves**

*(Radiotherapy, Brachytherapy)*

**Paris Dosimetry System:** For the Paris system dose calculations, dose rates in water at the central plane of a linear source are needed, where oblique filtration and attenuation and scattering in water are taken into account. Dose rate information for $^{192}\text{Ir}$ wires is available for instance in Dutreix et al. (1982) and Pierquin and Marinello (1997). The information is given both graphically and in tables. In the graphical presentation, the ‘escargot curve’, the central plane dose rate information is summarised in one single graph, showing the dose rate at varying distances for wires of unit linear source strength and different lengths.

Escargot means snail in French and the graphs have a shape that resembles a snail’s shell.

**Related Articles:** Paris system, Cross line curves


**EURATOM Treaty**

*(Radiation Protection)* The European Atomic Energy Community was established with a Treaty in 1957. This is one of the founding treaties establishing the European Union (EU). The Treaty, drafted in 1950, addressed mainly the topics considered important at that time, in particular regarding the field of nuclear power.

Under the provision of the EURATOM Treaty, the European Commission acquired the status of a supranational regulatory authority in the following fields: radiation protection, supply of nuclear fissile materials and nuclear safeguards. Of these the former is relevant to the operation of all facilities including also medical applications and hospitals.

The EURATOM Treaty makes little mention of criteria and norms to be respected during design or operation of facilities. Therefore these kind of regulatory activities have been developed under the responsibility of national authorities. International organisations and in particular the International Atomic Energy Agency (IAEA), have been able to achieve a certain level of standardisation. After years of work and several international conventions a ‘Culture’ of best practice has been established among the member states and therefore also the EU countries. The division between safety and health protection of the workers and public appears
Event-related design

(Magnetic Resonance) During a fMRI study, the subject undertakes a series of tasks known as a paradigm. These tasks can be arranged either in a block design (also known as a boxcar), or an event-related design. For an event-related design, stimuli (events) are presented not in ordered epochs, but briefly and randomly, so that the participant cannot predict if or when they will occur. Such experimental design allows the detection and analysis of the haemodynamic response function associated with individual events (in contrast with block designs, where there is simply a temporal integration of signal). By randomly mixing events of different types, it is possible to ensure designs, where there is simply a temporal integration of signal). By

A different section deals with the effects of non-ionising radiations. Event-related design has largely been fuelled by advances in the speed with which fMRI data can be acquired. By previous events (Figure E.42), it is possible to ensure that the response to any one event is not systematically influenced randomly mixing events of different types, it is possible to ensure that the response to any one event is not systematically influenced by previous events (Figure E.42).

Developments in event-related design have largely been fuelled by advances in the speed with which fMRI data can be acquired. Statistically, event related design is not as powerful as block design. Somehow ‘arbitrary’; there is in fact an effort towards a more comprehensive approach.

The protection of the health of the workers and public from the harmful effect of ionising radiations is the overall objective. Therefore the radiation protection unit, based in Luxembourg, deals with the following topics: (1) exposure of public; (2) occupational exposures; (3) emergency preparedness and response; (4) natural radiations; (5) medical exposures; (6) environmental monitoring and assessment and (7) education, information and training. A different section deals with the effects of non-ionising radiations.

In order to reach the objectives, it is necessary: (1) to propose and to implement community legislations; (2) to check the legal and operational implementation of community legislations; (3) to prepare basic safety standards; (4) to verify that member states perform their duties regarding obligatory monitoring of environmental radioactivity; (5) to provide a system of rapid information exchange in case of nuclear accidents; (6) to ensure implementation of maximum permitted levels in food and (7) to discuss radiation protection issues in meetings with the participation of independent experts.

Hyperlink: EURATOM: http://www.euratom.org

Event type in PET

(Nuclear Medicine) The annihilation coincidence detection (ACD) localises events along a line of response (LOR) when two coincidences are recorded in two opposite detectors within a specified coincidence timing window. If the detected photons originate from the same annihilation and have travelled without any scattering events, the event is referred to as a true coincidence (see Figure E.43a). The true coincidences are a good representation of the actual activity distribution. All coincidences recorded by the coincidence processor may not be true coincidences. Other coincidence events can also occur within the resolving time of the detector system. Two examples are shown in Figure E.43b and c.

As seen in Figure E.43c, a random coincidence occurs when two photons with unrelated annihilation points are detected within the coincidence timing window. The count rate for random coincidences for a detector pair is given by

$$R_{\text{random}} = \text{CTW} \times R_{\text{single,1}} \times R_{\text{single,2}}$$

(E.4)

where \( \text{CTW} \) is the coincidence timing window

$$R_{\text{single,1}}$$ and $$R_{\text{single,2}}$$ are the counting rates for the respective single channel detector

With greater administrated activity the quotient between random and true coincidences increases. As it follows from the preceding equation the count rate for random coincidences increases as the square of the activity present while the amount of true coincidences increases linearly. From the equation it also follows that the amount of random coincidences are proportional to the coincidence timing window. There is a lower limit for this window due to limitations in the electronics and time of flight considerations. The typical quotient for random to true coincidences in clinical PET scanners is 0.1–0.2 (brain imaging) and sometimes >1 for certain applications such as whole body imaging. Random coincidences appear fairly uniform across the field of view (FOV) and cause a loss of image contrast. It is also a source of error when quantifying the activity within the patient. One way to correct for random coincidences is to delay the coincidence time window. The delay should be much greater than the width of the window itself (e.g. 128 ns where width of window is 6 ns). The coincidences detected in this way are all random and will be approximately equal to the number of random coincidences detected in the prompt coincidence window. A correction can be made by subtracting the delayed coincidences. A third

Related Articles: fMRI (functional magnetic resonance imaging), Block design, Haemodynamic response function

category of events is when one or both photons are scattered before detection and as a result the LOR is misplaced. These coincidences are called scattered coincidences (Figure E.43b). The magnitude of the displacement is determined by the scatter angle and the location of the scatter event. The scatter event will most likely occur inside the patient but can also occur within components inside the scanner. The fraction of scattered events can in some cases (abdominal imaging) be as high as 60%–70%. This high fraction has three different causes. The first cause is that only one of the two scattering photons must scatter to produce a misplaced LoR. The second cause is that Compton scatter is the dominating interaction in scintillators at 511 eV and some incident photons might only deposit a small amount of energy. As a result the pulse height analysis window must be widened to include these interactions. The final cause is the low energy resolution in the LSO and BGO scintillators because of the low light output. It is therefore customary to use a wide pulse height analyser window. It is impossible to separate scattered events that originate in the body from those that originate inside the crystal and therefore the scatter method used in SPECT, e.g. a double-energy window is less successful. Today there are two main methods to deal with scatter in PET. The first one uses a transmission image and makes computational estimations of the scatter distribution for each projection. The second approach is to look at the scatter distribution in a projection profile (corrected for random coincidences) just outside the object. This profile is assumed to represent the distribution of scattered photons and the distribution is extrapolated over the entire projection using a cosine or Gaussian distribution.

Related Articles: PET, Beta decay

Event types in a scintillation camera
(Nuclear Medicine) There are four basic event types that can be registered in a gamma camera system. Only one of these event types provides correct positional information. The other event types cause degradation in spatial resolution (contrast) and errors in activity quantification. The first event type is called valid events. A valid event occurs when a photon is emitted near-parallel to the collimator holes, passes through the collimator and deposits all of its energy in the crystal with one photoelectric interaction. This is the event type that provides correct spatial information. Camera systems today are designed to increase the fraction of valid events by using crystal material with high atomic number, specially designed collimators with regard to photon energy and energy window discrimination.

The second event type is the detector scatter event. These events occur when a photon is emitted near-parallel to the collimator holes and Compton scattering in the crystal occurs. The scattered photon can either interact in a different location in the crystal or escape. In the latter case it is possible to discriminate events by analysing the pulse height, which is proportional to the energy deposited. If the energy deposited is not enough to reach the energy window the event is discarded. But if the scattered photon interacts in a different part of the crystal then the total amount of deposit energy may be within the energy window. The location of the event will therefore be placed somewhere in-between the two interactions. At high count rates, pile up can lead to a very prominent image artefact known as the pile up effect (separate article). Another type of detector scatter event occurs when photons scatter in the collimator before being registered. Such an event can be discriminated using an energy window. But if the energy resolution of the crystal is low some of these events might be registered as true events. These events will cause a loss of contrast.

The third event is the object scatter event. Such an event occurs when the photons are not emitted towards the collimator but scatter inside the object, pass through the collimator and deposit all energy at a single location. Such an event can be mis-located by several centimetres. The fraction of scattered events can be reduced by the use of an energy window, i.e. all events that do not register an energy deposition within the energy window are rejected. But if the scatter angle is less than 45° the loss of energy is minimal and the problem still remains. Object scattered events result in a low-spatial-frequency background that decreases the image contrast. Another source for scatter events is the collimator, i.e. when a photon Compton scatters in the collimator before being registered.

The fourth event is septal penetration. A photon emitted towards the collimator can penetrate the septal walls and interact in the crystal. This will cause a blurriness in the image since the system will assume that the event originated from a location perpendicular to the collimator face. This effect is more prominent when using high energy gamma-emitters. A way to minimise the number of septal penetration events is to use a collimator designed for the specific energy.

Events that are combinations of these simple events can also occur but are not as frequent.

Related Articles: Parallel-hole collimator, Pile up effect

Excess risk
(Radiation Protection) Excess risk is used to compare the risk in two different groups of people. For example, one group exposed to radiation for medical purposes, and the other group not exposed.

In this example, the groups may be compared to one another epidemiologically to see if belonging to that group exposed to the medical radiation increases their risk of developing certain diseases, particularly cancer. Any increase in the number of cancers detected in the exposed group, over and above that in the unexposed group may be termed the excess risk.

Excess risk should be analysed in comparison to total or absolute risk, and the relative risk compared to other factors.

Related Articles: Absolute risk, Relative risk

Excitation
(Magnetic Resonance) In physics, excitation is the elevation of an energy level over an arbitrary ground energy state. The lifetime of an excited system is usually short and determined by relaxation processes. In MR an excitation is obtained by transferring energy into the spinning nuclei using a radio frequency (RF) pulse. The length and amplitude of the excitation RF-pulse is chosen in a way (flip-angle) to produce net transverse magnetisation, which is measured afterwards. In MRI, after excitation data in k-space is measured. Furthermore, a number of excitations and subsequent measurements with the same parameter settings are used to average the data and sometimes called number of excitations (NEX).

Related Articles: Transverse magnetisation, Relaxation, Flip angle

Exit dose
(Radiotherapy) The exit dose D_{exit} is the absorbed dose delivered to a surface where the central axis of radiation emerges from the patient. At the patient exit side there is a lack of backscatter radiation because of a reduction in either the number of backscatter photons
or secondary backscatter electrons. These effects result in a build
down layer near the exit surface of the patient. The lack of backscatter
photons is dependent on the field size and increases with the patient
thickness and it is higher with lower photon beam energy. The lack
of photon backscatter influences a large region. The lack of second-
ary backscatter electrons influences the last millimetres of the patient
tissues in front of the exit surface. The position at which the exit dose
should be defined must be carefully evaluated. Even if the build-up
and build-down thicknesses are different they can be assumed as
equal for many measurements. $D_{exit}$ could be derived from measure-
ments with a detector placed at $d_{max}$ from the exit surface where $d_{max}$
is the build-up depth. The detector used in the exit dose determina-
tion should have a build-up thickness around its sensitive volume to
ensure complete electron backscatter. The same irradiation condi-
tions must be ensured in the case of the determination in a phantom
of the ratio between the dose to the detector and the patient exit dose.

Exit window

(Diagnostic Radiology) The term exit window is used to describe
the part of the x-ray tube glass envelope through which the useful
beam of x-rays exits the tube. At this place, the glass of the tube
envelope (usually borosilicate glass) is made thinner in order to
reduce the absorption of the useful beam. When the tube is placed
in the tube housing the exit window is adjusted to face the opening
of the housing. In x-ray tubes with metal housing the exit window is
made from metal with low absorption – usually beryllium ($Z = 4$) –
this is typical for mammographic x-ray tubes (1 mm is a common
thickness).

Related Articles: Glass envelope, X-ray tube

Expert systems

(Diagnostic Radiology) Expert systems are artificial intelligence
programs that attempt to reproduce or emulate the performance of
a human expert in a particular specialist problem solving task. In
order to simulate the performance of an expert a knowledge base
has to be formed as the system learns from the performance of a
specialist or subject matter expert. The expert system is then com-
pared to a human expert to verify its performance. An example of
such systems is the expert system that was used to simulate the per-
mance of a human expert in a particular specialist problem solving task. In

Exposure

(Radiation Protection) The quantity of $X$ or gamma radiation mea-
ured in terms of the ionisation produced in air is known as the
exposure.

Exposure is the quantity of $X$ or gamma radiation measured as
the sum of the electric charges ($Q$) on all the ions of one sign (+ve
or −ve) liberated and collected in air when all the secondary elec-
trons liberated by the photons in a small mass ($dm$) of air are com-
pletely stopped in air:

$$ X = \frac{dQ}{dm} $$

The historical unit of exposure is the Roentgen (R), while the SI
unit is Coulomb per kg:

$$ 1R = 2.58\times10^{-4} C/kg $$

Exposure does not include ionisation due to secondary photons
(e.g. Bremsstrahlung), therefore exposure can be considered to be
the ionisation equivalent of air kerma.

This is an important definition – ionisation chambers are used
for many applications for the actual physical measurement of
radiation output, e.g. from an x-ray or radiotherapy unit. The sub-
sequent conversion of that measurement into radiation dose units
(e.g. absorbed dose in air, air kerma, etc.) is a purely mathematical
exercise.

Related Articles: Roentgen (R), Absorbed dose, Air kerma,
Bremsstrahlung

Exposure counter

(Diagnostic Radiology) A device (or function) within an x-ray
machine to record the number of exposures produced.

Exposure point

(Diagnostic Radiology) Exposure point (or exposure number, or
L number) is an old measure used for adjusting the parameters of
the x-ray exposure used by radiographers in some countries. This
measure is analogous to similar measure used in film photogra-
phy. As per the exposure point system, introduced by Klaasen,
one exposure point is equal to the necessary change of x-ray expo-
sure (dose), which will produce exactly the same darkening of the
x-ray film, if the object thickness changes with 1 cm. Based on
exposure points radiographers had built exposure tables, with the
help of which they can easily calculate the optimal exposure for a
radiograph by simply measuring the thickness of the patient. The
introduction of AEC (automatic exposure control) has replaced
this system.

Exposure rate

(Diagnostic Radiology) The rate at which radiation is delivered to
a specified location (as mR/s). For example, the exposure rate pro-
based on measurement of radiation passing through the patient
Exposure time

(Diagnostic Radiology) Exposure time is the duration of the time
the x-radiation during a radiographic procedure (usually measured
in ms). It is either manually set by the radiographer or controlled by
the automatic exposure control (AEC) function of the equipment
based on measurement of radiation passing through the patient
and reaching the receptor. AEC is sometimes known as phototiming.

Relevant...
Exposure time
(Radiotherapy) This is the time for which the radiation field is applied. Sometimes it is called beam on time.

This term is used for radionuclide sources (Cobalt unit, brachytherapy).

For external beam radiotherapy (linear accelerators, x-ray orthovoltage and superficial machines), this concept has largely been replaced by monitor units. In other words, a number of monitor units is set up on the treatment machine instead of exposure time, in order to control the delivered dose.

Related Articles: Beam on time, Monitor unit

Exposure time limit
(Radiation Protection) Exposure time limits must be evaluated in the case of occupational working conditions with exposure to external ionising radiation and/or internal contamination. The evaluation should be made preventively, on the basis of the workload and of the work organisation, by the radiation protection officer/qualified expert. The exposure time allowed, for each worker, for each working condition should be as precise as possible, in order to take into account the characteristics and the risk related to the various working conditions. The aim is to be sure that the occupational workers are operating, with sufficient confidence and safety, within the occupational dose limits. Standard operational practices shall be indicated in order to minimise the risk and optimise the procedures. Training of personnel and routine verifications of the working conditions and related risks, for the workers, are an essential part of any safety assessment programs.


Extended field of view (FOV)
(Ultrasound) In current commercial ultrasound scanners, the FOV of the displayed image is dictated by the design of the transducer and its operation. For linear arrays, the width of the image is determined by the length of the array; for curvilinear arrays – by the length of the array and its radius of curvature – and in phased arrays – by the length of the array and sector angle.

Historically, extended-FOV images were available by combining images from an array of separate transducers or by storing several images from different transducer positions and using them for a composite image.

Location of the image position is key to this and was achieved in these systems by using multiple fixed transducers or by tracking the probe position. Optical and magnetic tracking of ultrasound probe position is still used for image registration in research applications, for example in correlating ultrasound images with 3-D MRI or CT neuro-imaging.

With increased computing power, several manufacturers offer extended-FOV imaging using the image itself as a reference and comparing features in subsequent images with the original (Figure E.44). By estimating displacement of the images and the velocity of the probe movement, extended views of tissue long the axis of the transducer can be constructed (Figure E.45).
Extended SSD treatment

(Radiotherapy) The term extended SSD treatment can be used in two situations. In one the SSD is set to a value beyond the nominal isocentre position of SSD = 100 cm to allow a larger field size to be used. The other situation is one where surface obliquity prevents a standard offset being used with an electron applicator, and so a larger SSD value must be used. For these situations it is important that the virtual source position is known and so the effective SSD can be found.

**Abbreviation:** SSD = Source to skin distance.

**Related Articles:** Effective source point, Virtual source position, Source-to-skin distance (SSD), Electron applicator, Electron oblique incidence, Effective SSD

Extension cylinder cone

(Diagnostic Radiology) A simple beam restrictor – a metal extender, similar to cylinder cone, but with an additional outer cylinder, that can move the restrictor closer to the radiographed area without change in the focus to image distance. See article on Beam restrictor.

**Related Articles:** Beam restrictor, Cone

Extent

(Nuclear Medicine) The polar map image can be used to determine the size and location of defects in a myocardial perfusion scan (Figure E.46). The extent of a myocardial perfusion abnormality is expressed as the number of pixels which fall below the normal limit as a percentage of the total number of pixels of the myocardium. Extent and severity are two important parameters for assessment and management of coronary artery disease (CAD).

**Related Article:** Severity

External beam irradiation

(Nuclear Medicine) Cyclotrons are used to produce radionuclides by accelerating charged particles on to a target.

One way to irradiate the target is to insert it directly into the particle beam path. This technique is called internal beam irradiation. Using this technique allows most of the accelerated particles to hit the target.

The second alternative is to extract the particle beam from the cyclotron and direct it on to an external target using either a stripping foil or an electrostatic deflector. This is called external beam irradiation. In general, external beam radiation is the preferred method. One of the disadvantages with external beam radiation is the low extraction efficiency. As much as 30% of the beam is not extracted, especially for cyclotrons using positively charged particles (positive ion cyclotron). Here electrostatic deflectors are used. The non-extracted beam irradiates and activates surrounding septa which leads to an unwanted radiation dose to cyclotron personnel. In modern negative ion cyclotrons, however, full extraction (close to 100%) is achieved with the use of a stripping foil.

**Related Articles:** Cyclotron, Internal beam irradiation, Stripping foil


External beam therapy

(Radiotherapy)

**Vocabulary in Radiotherapy:** Radiotherapy can be divided into two types of therapy, teletherapy and brachytherapy.

In teletherapy (from the Greek word tele which means ‘far off’), the sources are placed far from ‘the tumour’ (more than 5 cm).

In brachytherapy (from the Greek word brachy which means ‘short’), the sources are placed close to or inside ‘the tumour’ (less than 5 cm).

External beam therapy is in everyday speech said to be identical with teletherapy. It is to be noted though, that there are external beam techniques with very short source-skin distances, such as contact treatment with a contact x-ray machine, using a source-skin distance of 4 cm.

**External Beam Therapy Characteristics in General:** In external beam radiotherapy the radiation source is at a certain distance from the patient and the target within the patient is irradiated with an external radiation beam.

Most external beam radiotherapy is carried out with photon beams, some with electron beams and a very small fraction with more exotic particles such as protons, heavy ions or neutrons.

**Photon External Radiotherapy:** Photon external beams are mostly used in external beam radiotherapy. They are all characterised by the same physical parameters, but fall into various categories depending on their origin, means of production and energy. There are two origins of photon beams: gamma rays, which originate from radioactive nuclei, and x-rays, which originate in a target bombarded with energetic electrons. The x-rays from a target consist of Bremsstrahlung photons (Bremsstrahlung x-rays) and characteristic photons (characteristic x-rays). X-rays are produced either in an x-ray tube (superficial or orthovoltage x-rays) or in a linear accelerator (megavoltage x-rays).
External photon beam radiotherapy is usually carried out with more than one radiation beam in order to achieve a uniform dose distribution inside the target volume and as low as possible a dose in healthy tissues surrounding the target.

Modern photon beam radiotherapy is carried out with a variety of beam energies and field sizes under one of two set-up conventions: a constant source to surface distance (SSD) for all beams or constant source to axis distance (SAD) for all beams. In an SSD set-up, the distance from the source to the surface of the patient is kept constant for all beams and must be set up for every beam during irradiation, while for an SAD set-up the centre of the target volume (inside the patient) is placed at the machine isocentre once before the irradiation.

Clinical photon beam energies range from low (30–80 kVp) used for treating skin cancer and superficial structures, through medium (100–300 kVp), to megavoltage energies (60Co to 25 MV) used to treat deep-seated tumours (e.g. bladder, bowel, prostate, lung, brain). There is no sharp division between the various voltage ranges, and they can vary slightly in different documents.

Field sizes range from small circular fields used in radiosurgery, through standard rectangular and irregular fields, to very large fields used for total body irradiation (TBI).

External beam radiotherapy is usually carried out in fractions. The total prescribed dose is divided into fractions (normal fractionation is one fraction per day, 5 days a week, 2 Gy per fraction).

**Abbreviations:** SSD = Source to axis distance, SAD = Source to surface distance and TBI = Total body irradiation.

**Related Articles:** Deep therapy, Linear accelerator, Conformal radiotherapy, Photon beam, Electron beam, Heavy particle beams, Stereotactic radiosurgery, Radiotherapy, Intensity modulated radiation therapy, Cobalt unit, Orthovoltage, Isocentre, Fractionation


**External photoelectric effect**

*(Radiation Protection)* An interaction between photonic non-ionising radiation incident on the surface of a material (almost invariably a metal) in which the absorption of a photon in the material results in the excitation of an electron such that it is liberated from the surface of the material.

**Related Articles:** Photoelectric effect, Internal photoelectric effect

**Extinction cross section**

*(Ultrasound)* The extinction cross section (or total cross section) is defined as the time-averaged total absorbed and scattered power divided by the time averaged incident intensity. The unit is in square meters, as implied by the name cross section. Physically this cross section corresponds to the area of the incident wave that contains the amount of power that is absorbed and scattered by an object. Consequently, it is the sum of the scattering and absorption cross sections, defined analogously.

**Related Articles:** Scattering cross section, Absorption cross section, Differential scattering cross section

**Extracorporeal elimination**

*(Nuclear Medicine)* Extracorporeal elimination is an approach used to minimise the radiation dose to normal tissue in patients undergoing radioisotope therapy. The ideal radionuclide used for therapy would accumulate selectively to the tumour cells and spare normal tissue from most of the radiation dose. There are a few suggested approaches on how to minimise the radiation dose to normal tissue.

One of the methods suggested is the extracorporeal elimination. A compound that binds with high affinity to the excessive targeting agents in systematic circulation can be introduced subsequent to targeting agent injection. The compound binds to the targeting agent and speeds up the targeting agent clearance from the body, hence decreasing the radiation dose. One example of extracorporeal elimination is the extraction off excessive biotinylated and radionuclide labelled antibodies using an extracorporeal column with immobilised avidin.

**Related Articles:** Radionuclide uptake in tumour cells, Anti-idiotypic antibody technique


**Extraction fraction**

*(Nuclear Medicine)* Extraction fraction is the fraction of tracer in blood which is extracted in one pass through the vascular bed of a tissue. In terms of the tracer concentration in blood, it can be defined as the difference of between arterial and venous blood concentration divided by the concentration in arterial blood:

\[
E = \frac{C_{\text{arterial blood}} - C_{\text{venous blood}}}{C_{\text{arterial blood}}}
\]

The extraction fraction can also be thought of as the probability that a tracer molecule is retained in the tissue during a single pass.


**Extrafocal radiation**

*(Diagnostic Radiology)* The extrafocal radiation is a phenomenon in x-rays generation in the x-ray tube. Some of the accelerated thermal electrons are scattered back from the anode (without creating x-rays). These scattered electrons fall again on the anode (but outside the actual focal spot) and create x-rays outside the effective focal spot. This way the extrafocal radiation enlarges the effective focal spot and leads to blurring the radiography – see the figure in the article *Focal spot effective*. Some sources name this effect off-focus radiation. In stationary x-ray tubes the scattered electrons fall back on the anode stem (outside the target) and in this case the extrafocal radiation is called also stem radiation.

Various constructions of the x-ray tubes allow for minimising this effect, which otherwise can reach to 15%–20% of the useful radiation.

One possible construction minimising the extrafocal radiation is a metal ‘electron-capture’ hood, placed over the anode. This hood retains the scattered electrons. The anode hood is made of copper with an opening at the side of the cathode. At the path of the useful radiation the copper is replaced with Beryllium plate which can pass the x-rays with minimal absorption (Z = 4 for Be) – see the figure in the article *Stationary anode*.

**Related Articles:** Anode, Target, Stationary anode, Focal spot effective

**Extrapolated range of electrons**

*(Radiotherapy)* The extrapolated range of electrons (Z /χ) is found by using a plot of the central axis percentage depth dose curve (see *Figure E.47*). It is found by extrapolating the linear portion of the curve to the abscissa (x-axis, depth). The extrapolated range of electrons is sometimes also called the practical range, and while the two
methods for calculating these are different the typical values will tend to be similar. Therefore the rules for approximating the practical range are a good approximation for the extrapolated range, and so the practical range tends to be the more common figure quoted.

**Abbreviation:** PDD = Percentage depth dose.

**Related Articles:** Electron ranges, Electron maximum range, Electron therapeutic range, Electron practical range

### Extrapolation ionisation chamber

(Radiation Protection) In an extrapolation ionisation chamber some of its parameters (e.g. distance between electrodes (spacing) and thus volume) can be varied for measurement purposes. Then the measured values at different volumes/electrode spacing can be used by extrapolating the results to zero volume to determine the absorbed dose.

The measurement of the absorbed dose $D_m$ based on the Bragg–Gray principle:

$$D_m = W \times S_m \times P [\text{Gy}]$$

where

- $W$ is equal to the average energy loss per ion pair in the gas
- $S_m$ is a relative mass stopping power of the material to that of the gas
- $P$ a number of ion pairs formed in the gas per unit mass

The absorbed dose can be measured using a gas-filled ionisation chamber with the cavity small in comparison with the range of primary or secondary charged particles of measured radiation. The cavity is filled with gas, e.g. air, and the wall of the chamber is made from the solid material air-equivalent i.e. $S_m = 1$. The measurement of the dose in biological tissue is performed using so called tissue equivalent ion chambers. For the x-rays or gamma radiation of very low energies the measurement of the absorbed dose, e.g. in tissue, is realised with use of the extrapolation chamber because of a very small ion current in a tissue equivalent ion chamber. The extrapolation chamber consists of a pair of electrodes which spacing can be adjusted. The ion current per unit volume is measured for different distances between electrodes to permit extrapolation to zero chamber volume (zero spacing). The absorbed dose can be calculated from the extrapolated ion current using Bragg–Gray principle.

**Related Article:** Ionisation chamber


### Extrapolative dosimeters

(Radiation Protection) Radiation workers may receive high doses to specific parts of the body because of the work that they are undertaking. Common sites for high doses to occupational workers are hands and eyes. In medicine, radiologists or cardiologists performing interventional procedures under fluoroscopic control may receive hand and eye doses. Personnel dosimeters have been developed specifically for these sites.

The main requirement for hand and eye monitoring is to produce a dosimeter that is sufficiently small and flexible. The material which best suits this requirement is TLD. Personnel monitors for hands include rings and ‘finger stalls’, the latter used when it is the pulp of the finger that is most at risk of exposure, e.g. when injecting radioactive substances.

Tapes containing TLD are also available to be attached to glasses or directly to the forehead or neck when monitoring for eyes or thyroids is required.

**Related Articles:** Finger ring dosimeter, Integrating dosimeter, Thermoluminescent dosimeter (TLD)

### Extravascular

(Ultrasound)

See Comet tail

### Extremity coil

(Magnetic Resonance) An extremity coil is an RF coil used in imaging the knee, ankle, foot or wrist (Figure E.48). Most frequently the term ‘extremity coil’ is used with reference to a knee coil, and ‘knee coil’ and ‘extremity coil’ are often used interchangeably. Some extremity coils are designed either for knee or for foot/ankle imaging, whereas other designs combine both capabilities.

As volume coils, knee coils can be designed in a transmit/receive (T/R) configuration. Use of a T/R coil design avoids the need for RF excitation by the body coil and reduced SAR (specific absorption rate).

**Related Article:** RF coil
Falling-load generator
(Diagnostic Radiology) This type of x-ray generator is specially designed to produce short x-ray exposures (useful for radiographing moving organs). It does not use the normal way of producing the exposure with constant anode current (mA) but changes the anode current during the exposure (Figure F.1). This way the generator uses the maximal possible mA (following the maximal permissible mA from the tube loading chart). Applying the maximal mA allows the same mAs to be achieved for the minimal exposure time.

Figure F.1 gives an example presented at a set of typical tube load chart.

If an exposure with 120 kV and 10 mAs is required, it can only be achieved by applying 80 mA for 125 ms (80 mA × 0.125 s = 10 mAs). If higher mAs are used, the necessary time will go above the maximal permissible load curve for 120 kV (100 mA and 100 ms cross above the line).

If falling mode operation is used, initially the exposure begins with 200 mA for 10 ms (equal to 2 mAs), continues with 150 mA for 20 ms (equal to 3 mAs), and then continues with 100 mA for 50 ms (equal to 5 mAs). This way 2 + 3 + 5 = 10 mAs are delivered for 80 ms – i.e. 64% of the time of the first exposure during the operation, the x-ray tube load is always kept below the maximum permissible load.

Despite the shorter exposure time, these generators could lead to focal spot enlargement (hence decreased resolution). This is due to the quick heating of the anode during this high mA operation. Falling load generators are more expensive and usually work in radiographic systems with automatic exposure control. The production of more powerful x-ray tube gradually decreases the need of these x-ray generators.

Related Articles: High voltage generator, Tube rate chart

False negative (FN)
(General) The diagnosis of a disease using some kind of imaging modality, such as a scintillation camera, involves some kind of uncertainty in the decision of whether a patient has an abnormality or is normal. There can be four alternatives:

1. The diagnosis indicates a positive answer (disease) to a patient that has a disease.
2. The diagnosis indicates a positive answer (disease) to a patient that does not have a disease.
3. The diagnosis indicates a negative answer (no disease) to a patient that has a disease.
4. The diagnosis indicates a negative answer (no disease) to a patient that does not have a disease.

The preceding situations are often labelled as follows:

1 is the true positive (TP)
2 is the false positive (FP)
3 is the false negative (FN)
4 is the true negative (TN)

The fraction of true positive to all cases is called the sensitivity, TP/(TP + FN), and the fraction true negative to all cases is called specificity TN/(TN + FP). A ROC analysis determines the fractions of 1–4. The result is usually plotted as a ROC curve. The following figure shows such curves of two imaging modalities where modality B is better than modality A.

Related Articles: True positive, False positive, True negative
Far zone
(Ultrasound) See Diffraction

Farazd (General) The units of capacitance in the International System of Units (SI) are C V⁻¹ or farads (F). If a capacitor carries charge Q coulombs on either plate when the potential difference between its terminals is V volts, then its capacitance is Q/V farads.

Abbreviation: SI = International System of Units.
Related Article: Coulomb

Faraday cage
(Magnetic Resonance) A Faraday cage is a conductive enclosure designed to prevent the penetration of external electromagnetic radiation to a sensitive device housed within it. In MRI, the room housing the scanner is built as a Faraday cage. Without the cage, external sources of radio frequency (RF) radiation would interfere with the sensitive measurement of the weak nuclear magnetic resonances of the spins. The cage is constructed by lining the walls of the room with a continuous conductive surface made of sheets of copper, aluminium or steel. The conductive metal cage is not normally visible as the room is lined with a decorative finish. The cage continues through the observation window of the room as a conductive mesh set into the window glass. The door frame must be lined with conductive contact strips to ensure continuation of the conductive cage through the room door. Power and signal cables are routed to the MRI via a penetration panel which uses dedicated filters to prevent penetration of RF signal into the room. Any external tubing or fibres must be transferred into the room through conductive tubes called waveguides. The diameter and length of the waveguides are chosen to prevent transmission of RF signal with frequencies in a range that could interfere with imaging (Figure F.3).

Faraday shield
(Magnetic Resonance) The term Faraday shield refers, in the context of MRI, to the conductive enclosure of the entire examination room where the scanner is installed. It may also be denoted radio frequency (RF) shield since its purpose is to prevent electromagnetic radiation at the radio frequency range and particularly at the resonance frequency of the MR-scanner from passing in or out. Since the field strengths of clinical MR-scanners may typically be 0.2–3.0 T, the frequency range that needs to be heavily attenuated by the shield is (according to the Larmor equation) on the order of 10–130 MHz. The attenuation in this frequency range for a well-functioning shield can be 90 dB or better.

The shield is normally constructed using copper foil mounted on wooden sections. Stainless steel may also be used, but is more difficult to work with. The sections are put together at the inside of the walls, floor and roof of the scanner room and every part needs to be electrically connected to one solid cage covering the room. The entire cage also needs to be electrically insulated from the rest of the building. Special solutions such as frequency filters are required at spots where the shield needs to be penetrated. Therefore, there is a penetration panel where for example cables needed for power and signal cables are routed to the scanner via a penetration panel which uses dedicated filters to prevent penetration of RF signal into the room. Any external tubing or fibres must be transferred into the room through conductive tubes called waveguides. The diameter and length of the waveguides are chosen to prevent transmission of RF signal with frequencies in a range that could interfere with imaging (Figure F.3).

If any part of the shield is damaged, for example by running a nail through it, or if the door is left open, the attenuation no longer works and image artefacts may occur due to radio signals that the MR-scanner picks up from the outside environment.
Farmer chamber

(Readiotherapy) The Farmer chamber is used to determine radiation dose by measuring the charge accumulated between two electrodes when the chamber is irradiated. One electrode is the cylindrical graphite outer electrode and the other is a central aluminium electrode, see Figure F.4a. The chamber dimensions are shown in Table F.1.

Farmer chambers may be supplied with a build-up cap (Figure F.4b). The chamber is placed inside this to achieve CPE when irradiated in air in a cobalt 60 beam; however, the build-up cap is not used when measuring doses routinely in the clinic.

This device can be used for relative or absolute dose measurements, and because the chamber is open to the atmosphere, a temperature and pressure correction has to be made to readings where absolute measurements are required.


Fast field echo (FFE)

(Magnetic Resonance) FFE refers to a gradient echo generated at an echo time after the radiofrequency excitation pulse by applying a bipolar gradient pulse in the readout (frequency) direction. Since the echo is formed during the free induction decay (FID), the echo signal exhibits a $T_2^*$ weighting. The term FFE can be used interchangeably with the term gradient echo. See related article which also includes a short description of the basic gradient echo pulse sequence.

Related Articles: Fast low angle shot (FLASH), Free induction decay (FID), Gradient echo

Fast Fourier transform

(General) The FFT is a powerful efficient algorithm for computing the discrete Fourier transform (DFT). For $N$ data points, it reduces the number of computations from the order of $N^2$ to the order of $N \log_2(N)$. If the data are 1024 long, the saving is a factor of 100. An FFT produces exactly the same result as evaluating the DFT directly.

Most FFT algorithms operate in optimal efficiency when $N$ can be factorised easily, for example if $N$ is a power of 2. The Cooley–Tukey FFT algorithm is such an example.
**Summary:**

- **Fast imaging with steady state precession (FISP)**
  - **SSFP** belongs to the family of steady state free precession (SSFP) sequences. This is a gradient echo sequence that combines short repetition times and low flip angles to establish a steady state of the transverse magnetisation. There are several variants in the family of steady state free precession depending upon how the gradients are designed to be balanced in one, two or all three gradient directions.
  - **Fast imaging with steady state precession (FISP)** (Magnetic Resonance) The free induction decay steady precession (FISP) sequence belongs to the family of steady state free precession (SSFP) sequences. This is a gradient echo sequence that combines short repetition times and low flip angles to establish a steady state of the transverse magnetisation. There are several variants in the family of steady state free precession depending upon how the gradients are designed to be balanced in one, two or all three gradient directions.

**Abbreviations:**
- FFT = Fast Fourier transform and DFT = Discrete Fourier transform.
- Related Articles: Fast Fourier transform, Discrete Fourier transform


**Fast imaging with steady state precession (FISP)**

**Steady state free precession (SSFP)**

**Further Reading:** Scheffler, K. 1999. A pictorial description of steady-states in rapid echoplanar MR imaging techniques. Scheffler, K. and J. Hennig. 2003. Is trueFISP a gradient-echo sequence or a spin-echo sequence?

**Related Articles:** Fast Fourier transform, Discrete Fourier transform

**Further Reading:** Scheffler, K. and J. Hennig. 2003. Is trueFISP a gradient-echo sequence or a spin-echo sequence?

**Figure F.5a** shows the timings of a FISP sequence, i.e. a SSFP sequence balanced in the phase encode direction but unbalanced along the readout-direction. The difference between a FISP sequence and a FLASH sequence is the balanced gradients in the phase encode direction and the lack of spoiler gradients or spoiling RF pulses. For short TRs and large flip angles, the SSFP-FID sequence provides images of a mixed $T_1/T_2$ contrast, using short TRs, low flip angles and short TEs; the contrast is proton density (PD) weighted, but with a long TE, the sequence gives $T_2^*$-weighted images.

**Figure F.5b** shows a true FISP sequence, in which all three gradient directions are balanced. The image contrast of such a balanced SSFP sequence is related to the $T_1/T_2$ ratio. Therefore, tissues with a high ratio, such as bile, blood and fat appear bright. The sequence allows ultra-fast imaging in all areas of the body and is regularly used in cardiac MRI, MRA, foetal imaging in utero and for MR-guided interventional procedures.

**Acronyms for the SSFP-FID sequences are as follows:**
- Balanced in one direction: GRE /GRASS (General Electric), FFE (Philips) and FISP (Siemens).
- Balanced in three directions: FIESTA (General Electric), balanced FFE (Philips) and true-FISP (Siemens).

**Related Articles:** Fast low angle shot (FLASH), Flip angle, Steady state free precession (SSFP)


**Fast spin echo (FSE)**

(Fast Imaging using Steady state Precession) FSE is an MRI pulse sequence characterised by the encoding of multiple k-space lines in a train of spin echoes (Figure F.6). For example, if three echoes are sampled after each 90° pulse, the total acquisition time will be decreased by a so-called ‘turbo factor’ of three compared to a single-echo spin echo sequence with identical parameters. The number of echoes is also known as the echo train length, ETL.

The time period between the excitation pulse and the acquisition of the central k-space line is denoted the effective echo time, $T_{EE}$, and this parameter can be changed by adjusting the echo spacing (the time between echoes) and by ordering of the k-space line sampling strategy. Since the echoes associated with the centre of k-space determine the contrast in the images, $T_{EE}$ gives $T_1$ – or PD weighting (depending upon the choice of repetition time) while high $T_{EE}$ (late encoding of the echoes in the centre of k-space) in combination with a long repetition time will lead to $T_2$-weighted images.

Since a high number of powerful 180° radiofrequency pulses are applied, FSE sequences tend to have a high specific absorption ratio (SAR). SAR limits may be met by reducing this flip angle and using hyper-echo techniques.

Other names for variants of this sequence are turbo spin echo (TSE), rapid acquisition relaxation enhancement (RARE) and half-Fourier single-shot turbo spin echo (HASTE).

**Related Articles:** Echo train length, Half acquisition single-shot turbo spin echo (HASTE), Rapid acquisition relaxation enhancement (RARE), Turbo spin echo (TSE)

**Figure F.5** Schematic illustration of (a) a partially refocused SSFP-FID sequence (FISP) and (b) a fully refocused SSFP-FID (true-FISP) sequence.

**Abbreviations:**
- $T_1$ = relaxation time
- $T_2$ = transverse relaxation time
- $T_2^*$ = transverse relaxation time (long TE)
- $T_{EE}$ = effective echo time
- $T_{RF}$ = repetition time
- $T_{TE}$ = echo time
- $n$ = number of echoes
- SAR = specific absorption ratio
- ETL = echo train length
- $E$ = an MRI pulse sequence characterised by the encoding of multiple k-space lines in a train of spin echoes, such as turbo spin echo (TSE), rapid acquisition relaxation enhancement (RARE) and half-Fourier single-shot turbo spin echo (HASTE).

**Related Articles:**
- Fast field echo
- Flip angle
- Gradient echo

**Further Reading:**
Fast timing techniques for single-channel analysers

(Radiation Protection) A timing single-channel analyzer marks the arrival time of detected pulses called an event time. The time resolution depends on the detector (the best is for the fastest detectors, e.g. scintillation detector) and on the ratio of maximum to minimum pulse height, i.e. dynamic range of pulses. However, it is limited by the following factors:

a. Jitter for constant amplitude of input pulses caused by pulse statistical fluctuations and input noise
b. Amplitude walk effect time slewing for variable amplitudes of input pulses

Both the factors are shown in Figure F.7 in leading edge triggering. The leading edge (LE) triggering is the easiest direct time pick-off method using a fixed discrimination level (threshold) to mark out the time at which a pulse crosses it.

Besides the leading edge triggering, there are other time pick-off methods as zero crossover and fast crossover timing, constant fraction (CF) timing, amplitude and rise time compensated (ARC) timing (e.g. for germanium detectors where the rise time variations are large), extrapolated leading edge timing (ELET) and first photoelectron timing (FPET) for scintillation detectors. The choice of the method depends on the pulse amplitude range (narrow or wide) and its shape as well as its rise time.

The best timing is provided for detectors having fast and non-variable rise time.

The timing single channel analysers (SCAs) are used in coincidence systems (e.g. events detected by two PM tubes occurred simultaneously) or in time spectroscopy (time interval measurement) in the range of $10^{-6}$–$10^{-12}$s. The time interval $\Delta t$ measurement may be performed with the time-to-amplitude converter (TAC). This device produces an output pulse with an amplitude proportional to $\Delta t$ between input start and stop pulses. The amplitude distribution may be measured by a SCA or multi-channel analyzer (MCA). The example of a coincidence measuring system and time spectroscopy is presented in Figure F.8.


Fat

(General) The term fat refers to lipids that are solid at room temperature. Fats are hydrophobic triglyceride molecules composed of one glycerol and three fatty acid molecules. Fats can either be saturated, where the fatty acid contains no double bonds, or unsaturated, where the fatty acid molecule does contain double bonds.

In the human body, all excess carbohydrates, proteins, fats and oils are stored in adipose tissue. Fat has twice the energy storage capacity per kilogram compared to carbohydrate.

Properties of Fat Based Tissues for Medical Imaging: Fat-based tissues are visible on both x-ray and magnetic resonance images. They have short $T_1$s and $T_2$s of between 100–150ms and 10–100ms, respectively. The CT number of fat-based tissues is about 120 Hounsfield units.
**Fat nulling**

*Magnetic Resonance* See Fat suppression

**Fat saturation (FATSAT)**

*Magnetic Resonance* Fat saturation, commonly referred to as FATSAT, selectively saturates lipid (fat) prior to an excitation RF-pulse. This technique employs a frequency-selective RF-prepulse and a subsequent spoiler gradient that saturates the fat resonances while leaving those of the water unaffected. This results in a negligible signal from the lipid protons in the images. Since this technique relies on frequency-selective RF-prepulses prior to the imaging/spectroscopy sequence, the fat saturation technique can be used with almost any pulse sequence.

Effects of magnetic field inhomogeneities can, however, result in a poor suppression due to the shift in the lipid resonance frequency. For example, if the frequency of the lipid protons shifts due to an alteration in the local magnetic field and this results in an unwanted suppression of off-resonance water signal.

**Related Articles:** CHEMSAT (chemical selective saturation), Fat suppression

**Fat suppression**

*Magnetic Resonance* Fat suppression can be achieved by three methods: fat saturation, inversion recovery imaging and opposed phase imaging. Each of the three methods has advantages and disadvantages, which will be described in the following.

Fat saturation is achieved by applying a frequency selective RF-pulse and a subsequent spoiling gradient, which suppresses the fat resonances while leaving the water signal unaffected. Fat saturation can be used with almost any pulse sequence since it is performed as a magnetisation preparation experiment before the excitation RF-pulse of the imaging sequence. Since fat saturation relies on a frequency-selective saturation pulse, effects of magnetic field inhomogeneities will result in a poor suppression due to the shift in lipid resonance frequency and an unwanted suppression of water resonances.

Short TI inversion recovery (STIR) is based on the application of an inversion RF-pulse before the imaging sequence. Since the $T_1$ relaxation time of fat is shorter than for water, the longitudinal magnetisation of fat will recover faster than that of water. The excitation RF-pulse of the imaging sequence is applied at an inversion time ($T_i$) when the longitudinal magnetisation of fat crosses zero ensuring suppression of fat signal. Usually, this occurs at 0.69 times the $T_1$ of fat. For example, the optimal for fat suppression at 1.5T is $T_i = 130–170$ ms. The STIR pulse sequence is the only method that is insensitive to inhomogeneities and can be used with low-field-strength magnets. However, the method is non-specific since water with $T_1$ similar to that of fat also will be suppressed. The inversion recovery preparation introduces an additional $T_1$ contrast, which can be suboptimal if a PD or $T_2$-weighted contrast is required. In addition, the signal-to-noise ratio is also reduced since the longitudinal magnetisation of the water will not have fully recovered at the time of the excitation.

The opposed phase technique is based on the different resonance frequencies between water and fat signal, which results in phase differences at different echo-times in gradient echo imaging. Usually, images at two different echo-times are acquired in which water and fat signals will interfere destructively (opposed phase) and constructively (in phase). At 1.5T, the opposed phase occurs at TE = 2.26 ms, whereas in phase at TE = 4.52 ms. The in-phase image ($I_{ip}$) is the sum of the fat and water signal, i.e. $I_{ip} = I_{water} + I_{fat}$, whereas the opposed-phase image ($I_{opp}$) is the difference image, i.e. $I_{opp} = I_{water} - I_{fat}$. A fully fat-suppressed image can be generated from two images with different TE as $I_{water} = (I_{ip} + I_{opp})/2$. If only one image is acquired, $I_{opp}$ can be used as a fat-suppressed image assuming the same amount of water and fat signal in each voxel. The technique is simple and fast, but the one-image method also affects the water signal, while the difference method with two images is sensitive to field inhomogeneities. Recently, techniques were proposed that can compensate for field inhomogeneities by acquiring three images at different echo times.

A fourth alternative to avoid the fat signal is to only excite the water using a frequency selective excitation pulse, but this is not a fat suppression method per se.

**Related Articles:** Fat saturation, Inversion recovery, Short tau inversion recovery (STIR)


**FATSAT (fat saturation)**

*Magnetic Resonance* See Fat saturation (FATSAT)
Feature extraction
(General) Feature extraction refers to the process whereby a set of measured features within an object is compared to established criteria in order to classify the object.

Feedback
(General) Feedback describes the arrangement where part of the output is fed back to the input to achieve automatic self-regulation of an electrical, mechanical or biological system. In automation, the use of feedback is fundamental control mechanism for machinery. Feedbacks are also widely used in electronic control systems.

In negative feedback systems, if the output signal increases, the feedback circuit reduces the signal in the input (and vice versa), thus stabilising the system. For example, in voltage and current regulators, part of the output is used as a control input, providing self-regulation.

In positive feedback systems, if the output signal increases, the feedback circuit increases the signal in the input, this way creating a generator of oscillations. Usually such systems have internal sensor for overloading, which interrupts the oscillations.

Feedback control systems of great complexity also exist in living organisms.

Ferromagnetic materials
(Magnetic Resonance) Like the mass and the electrical charge of a particular element or material, magnetism is a fundamental property of matter. All materials interact in some form with an external magnetic field. The different types of magnetic materials are usually classified in terms of their susceptibility or permeability. The permeability is defined as $\mu = \frac{B}{H}$, and the susceptibility is defined as $\chi = \frac{\mu - 1}{\mu}$, where $B$ is the magnetic induction, $H$ the magnetic field and $M$ the magnetisation, that is the magnetic dipole density. The apparent magnetisation of an atom is given by

$$\overline{M} = \chi \overline{H}, \; \text{thus} \; B = \mu_0 (\overline{H} + \overline{M}) = \mu_0 (1 + \chi) \overline{H}$$

where $\mu_0 = 4\pi \times 10^{-7}$ H m$^{-1}$ is the permeability of free space (in SI units), which is a universal constant.

The magnetic susceptibility of a material as detected by NMR is related to the ability of an external magnetic field to affect nuclei of a particular atom. This is strongly related to the electron configurations of that atom. For example, the nucleus of an atom which is surrounded by paired electrons is less affected by an external magnetic field than a nucleus with unpaired electrons. Materials having a negative value of susceptibility are called diamagnetic, when the susceptibility has a value between 0 and 1 materials are called paramagnetic and ferromagnetic if the susceptibility is greater than one.

Paramagnetism: Unpaired electrons in paramagnetic materials induce small magnetic moments. With no external magnetic field, these magnetic moments have a random orientation, and, therefore, they cancel each other out. In the presence of an external magnetic field, paramagnetic materials align with the direction of the field and so the magnetic moments add together, resulting in a local increase of the magnetic field.

Diamagnetism: Diamagnetic materials present a zero net magnetic moment with no external magnetic field. Diamagnetic materials present no elementary magnetic dipoles. This is because the electron currents caused by their motions add to zero. When an external magnetic field is applied, small magnetic moments are induced, which counteract the applied field (Lenz’ law). This determines a subsequent slight decrease in magnetic field within the sample. Diamagnetic materials have negative magnetic susceptibilities and are therefore slightly repelled by the magnetic field. Diamagnetic effects appear in all materials. However in a material which possesses both diamagnetic and paramagnetic properties, the positive paramagnetic effect exceeds the negative diamagnetic effect so the material appears paramagnetic.

Ferromagnetism: If a ferromagnetic material is in the presence of an external magnetic field, the results are strong alignment and attraction. The material retains its magnetisation even when the external magnetic field has been removed, and, therefore, they are permanently magnetised and subsequently become permanent magnets. The magnetic field in permanent magnets can be hundreds or even thousands of times greater than the applied external magnetic field.

In Figure F.9, a qualitative representation of lines of forces of the magnetic induction is shown for the paramagnetic, diamagnetic and ferromagnetic substances.
Ferromagnetism

In ferromagnetic substances, the atomic dipoles tend to align in the same direction over regions or domains containing millions of atoms forming a small magnetic dipole. The interactions between a ferromagnetic object and an external magnetic field may have two effects. If the net dipole forms an angle to the main magnetic field, it tends to align to the main field direction under the action of a torque. The object may also experience a force along the gradient of the magnetic field. Both torque and direct attraction can affect any ferromagnetic object inside the patient as well as outside in presence of an external magnetic field. A metallic cylinder, 3 mm in diameter and 75 mm long with a mass of 4.2 g, which forms an angle of 45° to a field strength of 1 T, requires a force of around 6000 N (600 g) applied at each end to prevent a twist. If the cylinder is free to move towards the magnet, it will achieve a final velocity of 17 m s⁻¹. The velocity is irrespective of the shape of the object and depends only on object mass.

Ferromagnetism

(Magnetic Resonance) The spin of the electron results in a magnetic dipole moment that creates a magnetic field. In atoms with unpaired spins, a net magnetic moment exists even in the absence of an external field. Ferromagnetic materials contain many atoms with unpaired spins; these tiny magnetic dipoles are aligned parallel to each other within small regions of the material to form areas of stronger magnetism. Compared to a paramagnetic material, a permanent magnetisation exists in the ferromagnetic material after the externally applied field is removed. Examples of ferromagnetic materials are iron (26), cobalt (27) and nickel (28). The numbers in the parenthesis are the atomic number of the material.

If a ferromagnetic object (e.g. an implant) is placed in an MRI scanner, the object will distort the homogeneity of the main magnetic field and cause susceptibility artefacts in the images.

Related Articles: Susceptibility, Paramagnetism

Ferrous sulphate dosimetry

(Radiotherapy) See Fricke dosimeter

F-factor

(Diagnostic Radiology) The F-factor is a conversion factor in radiology between the exposure defined as the amount of ionisation in air and the absorbed dose in tissue caused by the x-ray exposure.

The relationship between ionisation in air and in tissues is not fixed but depends on the type and energy of the radiation and on the type of tissue involved:

\[ F\text{-factor} = \frac{\text{Absorbed dose}}{\text{Exposure}} \]

In the United States, the traditional units used are:

- Ionisation in air: Roentgen (R) \( (1 \text{ R} = 2.58 \times 10^{-4} \text{ C kg}^{-1}) \)
- Absorbed dose: rad \( (1 \text{ rad} = 0.01 \text{ Gy}) \)

In these units, the F-factor for diagnostic x-rays varies between approximately 1 for soft tissues and up to 4 for bone:

- Most other countries use the MKS units:
  - Ionisation in air: Coulombs per kilogram \( (\text{C kg}^{-1}) \)
  - Absorbed dose: Joules/kilogram \( (\text{Gy}) \)

In this case, the MKS F-factor varies between approximately 40 (soft tissues) and 160 (bone).

Further Reading: Bushberg et al. 2002. The Essential Physics of Medical Imaging, Lippincott Williams & Wilkins, Philadelphia, PA.

Hyperlink: http://www.osha.gov/SLTC/radiationionizing/introtoionizing/ionizinghandout.html

FFE (fast field echo)

(Magnetic Resonance) See Fast field echo (FFE)

FFT (fast Fourier transform)

(General) See Fast Fourier transform (FFT)

Fibre optics

(General) An optical fibre is a specialised, flexible strand of glass or plastic used to convey light energy between two points.

An optical fibre consists of a high refractive index core coated with a lower refractive index cladding. Due to the difference in refractive indices between the two materials, light travelling in the core experiences total internal reflection at the boundary with the cladding and is transmitted along the length of the fibre. This geometrical optics analysis is no longer valid where the fibre core diameter is in the order of microns and fibre transmission must be explained by reference to the fibre as a waveguide. This analysis shows that where the fibre diameter approaches the wavelength of the light transmitted, some of the light (the evanescent wave) actually travels along the fibre through the cladding.

Fibre optics are used primarily for data transmission, where bursts of light transmitted down the fibre convey digital information between an electronic transmitter and receiver. At the transmitter, the data to be transmitted are converted into light pulses, typically using a miniaturised light emitting diode (LED) or laser diode. The light is coupled into the fibre for transmission to the receiver. At the receiver, the pulses are reconverted into electronic signals using a photodiode coupled to the fibre. In general telecommunications, optical fibres can support higher data rates than conventional copper wire connections.

In an optical endoscope, a bundle of optical fibres is used to convey an image from the distal tip of the endoscope to the eyepiece. For optical image transmission, the fibre bundle must be ‘coherent’, i.e. every constituent fibre must retain the same relative position within a cross section at the distal and proximal ends of the bundle. Fibre bundles are also used to deliver illumination from a source to a distant target. Bundles used for illumination need not be coherent.

In the MRI environment, fibre optics finds application where the use of conventional conductive wires for data transmission and signal acquisition is limited by RF interference and RF heating risks. For example, in a conventional pulse oximeter, a conductive cable connects the finger clip sensor placed on the patient to the pulse oximeter display. The finger clip houses a LED and photodiode and transmits light through the finger. Analysis of the light transmitted generates a pulse waveform and a value of % oxygen saturation. In an MRI compatible pulse oximeter, fibre optics is used to transmit and collect light to and from the finger clip, avoiding the use of a conductive cable near the MRI bore. Data collection from patient monitoring systems placed in MRI scan rooms to remote display consoles is typically via fibre optics, as is control of contrast injectors.

Fibre tracking

(Magnetic Resonance) Fibre tracking is a method for constructing and visualising axonal tracts based on anisotropic diffusion (e.g. in cerebral white matter and muscles). See Tractography for further details.

Related Article: Tractography
Fick's method
(Nuclear Medicine) Fick's method is a technique for determining consumption of a substance by an organ. It is calculated from the product of the arteriovenous concentration difference of the substance and blood flow. Mass is preserved, and, following that, the mass of a substance entering an organ minus the mass of the substance leaving the organ equals the mass of substance retained in the organ.

Related Article: Fick's principle

Fick's principle
(Nuclear Medicine) The Fick's principle states that the rate of change of tracer in an organ or tissue is equal to the difference between the amount of tracer arriving (usually via arterial blood) and the amount leaving (via venous drainage) the organ or tissue. If \( C_a \) and \( C_v \) denote concentration of tracer in arterial and venous blood, then the amount of tracer \( M_t \) in the tissue can be described as

\[
\frac{dM_t}{dt} = QC_a - QC_v - Y
\]

where

- \( Q \) is the blood flow
- \( Y \) is the rate of loss through excretion

If \( Y \) is zero, there is a simple relationship between the rate of change of \( M_t \) and the extraction fraction \( E \) (see related article):

\[
\frac{dM_t}{dt} = QC_a - QC_v = Q(C_a - C_v) = C_a(C_a - C_v) = C_aQE
\]

Related Article: Extraction fraction

FID (free induction decay)
(Magnetic Resonance) See Free induction decay (FID)

Field coverage
(Radiotherapy) This term relates to the size of the treatment field being sufficiently large to provide adequate uniform dose coverage across the required planning target volume from the beam's eye view. One of the aims of planning is to ensure that a sufficiently high (typically 95%) isodose line completely encompasses the target volume, and so the planner must design a plan to provide this.

Related Articles: Target dose distribution, Isodose surface, Planning target volume, Treated volume

Field echo
(Magnetic Resonance) See Gradient echo (GE)

Field emission display (FED)
(Diagnostic Radiology) FED is a form of flat panel display and may be compared to the more common LED and LCD displays found in modern television and computer monitors.

It is based on a two dimensional array of minute cathode ray tubes (CRT), each with an evacuated space containing an extremely small pointed cold electron emitter, a fine conducting grid and a coloured phosphor anode (in colour displays, one CRT unit forms a single pixel of one colour).

Unlike normal CRTs, no heater element is needed to cause the electrons to be emitted by the cathode. Rather, a positive voltage applied to each grid causes electrons to be emitted and pass through, where they are further accelerated towards the phosphor-coated anode by a large positive anode potential, where their energy is converted to visible light.

The non-linear current/voltage property of these mini-emitters allows for easy pixel addressing using a 2D grid of conductors running across and down the screen. Each pixel is selected by powering the appropriate column and row simultaneously, whilst the brightness of each pixel is controlled by varying the time for which the voltage is applied.

Despite their requirement for a vacuum within each cell, problems with practical cathodes, and the inefficiency of energy conversion to light, they have been shown to promise much higher power efficiency than present LCD displays.

They presently remain in the development/prototype stage.

Related Article: Cathode ray tube
Hyperlink: http://en.wikipedia.org/wiki/Field_emission_display

Field margin
(Radiotherapy) When planning radiotherapy treatments, field margins are often added to each beam's eye view projection of the planning target volume (PTV), to account for the finite beam penumbra. Typical values would be 0.5 cm; however, the field margin depends on the beam arrangement. For example, in axial coplanar treatments with multiple fields, large field margins are needed for definition of the superior and inferior field borders to avoid under-dosages in the equivalent areas of the PTV, since these areas do not have full contribution from other beams. In this case, the superior and inferior field margins can even be double the standard field margin.

Abbreviation: PTV = Planning target volume.

Related Articles: Penumbra, Planning target volume (PTV)

Field of view (FOV)
(Diagnostic Radiology) FOV of a CT scanner refers to the diameter of the circular area over which CT data are acquired (scan field of view [SFOV]) or over which data are reconstructed (reconstruction field of view [RFOV]).

The maximum SFOV of a clinical CT scanner is generally on the order of 50 cm (Figure F.10).

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**Figure F.10** Diagram demonstrating (a) maximum SFOV and (b) head SFOV. (Graphs courtesy of ImPACT, UK, www.impactscan.org)
Field of view (FOV)  
(Nuclear Medicine) FOV of a scintillation camera system is basically determined by the size of the scintillation detector and the depicted image. More contiguous factors that determine the FOV are ring size (PET), collimator design (scintillation camera), edge packing effects (scintillation camera) and the detector surface area (all detector systems).

Since parts of the FOV are unsuited for activation, quantification or imaging due to a number of reasons (edge packing, etc.), it is more relevant to discuss the useful field of view (UFOV) and the central field of view (cFOV).

**Related Articles:** Central field of view (cFOV), Useful field of view (UFOV)

Field strength  
(Magnetic Resonance) The strength of a field is the magnitude of its vector value. Two physical quantities may be referred to as the magnetic field: the magnetic field \( H \) and the magnetic induction \( B \) (also called magnetic flux density). A magnetic field can either be produced by electrical current or by a permanent magnet. The magnetic field generated by a current can be calculated from the Biot–Savart law or Ampère’s law. When a magnetic field has been generated in a medium, the response of the medium is its magnetic induction \( B \). The relation between magnetic induction and magnetic field is given by the equation \( B = \mu H \), where \( \mu \) is the permeability of the medium (H m\(^{-1}\)). Although the term ‘magnetic field’ was historically reserved for \( H \), the magnetic induction \( B \) is now understood to be the more fundamental entity and most modern writers refer \( B \) as the magnetic field. The magnetic induction (or magnetic field) \( B \) is measured in Tesla (T). In MRI, the strength of the main magnetic field varies from 0.2 to 3.0 T in clinical practice; whereas, in research, magnets with values of 7 T to higher than 11 T are used. Usually, the main magnetic field is generated by superconducting electromagnets. The main magnetic field determines the resonance frequency (called the Larmor frequency) and the net magnetisation, which is higher for higher fields. Furthermore, the \( T_1 \) relaxation time strongly depends on the magnetic field strength.

**Related Articles:** Larmor frequency, Net magnetisation, Magnets

Field uniformity  
(Magnetic Resonance) There are two types of field in an MRI scanner, the main static magnetic field and the RF field.

The main static magnetic field is referred to as \( B_0 \) and typically ranges from 0.1 to 3.0 T for clinical use. The uniformity of the magnetic field is an important criterion of the quality of the magnet as non-uniformities can lead to image artefacts. Magnets are, therefore, constructed to be as homogeneous as possible, especially at the isocentre.

The RF field (also known as \( B_1 \)) uniformity is affected by two main factors, the first is the interaction between the RF field and the object being imaged and the second is the inherent inhomogeneities of the RF coils used in the system. A uniform RF field is not always required, depending upon the area of the body that is being imaged. Structures close to the surface of the body maybe imaged using a surface coil, which is not uniform, while imaging of internal structures may use a more uniform RF coil design such as a birdcage coil.

**Abbreviation:** RF = Radiofrequency.

**Related Articles:** \( B_0 \) homogeneity, \( B_1 \) inhomogeneity, RF uniformity

Field verification  
(Radiotherapy) External beam radiotherapy is delivered as a set of treatment fields. It is recommended to verify each field independently. This may consist of two stages: verification prior to treatment and on-treatment verification. The verification is done by comparison of the shapes and the positions of the planned and treated fields.

**Field Verification Prior to Treatment:** This is often particularly useful for complex treatment such as intensity-modulated radiotherapy (IMRT) with dynamically moving multileaf collimator (MLC) leaves. Field shapes may be verified in comparison with a template. The intensity distribution of an IMRT field may be verified using integrated imaging with film or a digital imaging device such as an electronic portal imaging device (EPID).

**Field Verification on Treatment:** This generally involves imaging the position of the patient’s anatomy relative to the treatment field using film or a digital imaging system such as an EPID.

**Abbreviations:** IMRT = Intensity-modulated radiotherapy, MLC = Multileaf collimator and EPID = Electronic portal imaging device.

**Related Articles:** Radiotherapy verification, Portal film, Portal radiograph, Electronic portal imaging, Intensity-modulated radiotherapy

Field weight  
(Radiotherapy) See Beam weight

Field-effect transistor (FET)  
(General) Field effect transistors (FET) are widely used for amplification of digital or analogue weak signals, DC switches or as oscillators (Figure F.11).

In the FET, the current flows along a semiconductor path between the source and the drain electrodes. Its diameter defines FET conductivity.

There are two types of FET–junction FET (JFET) and the metal-oxide-semiconductor FET (MOSFET) (Figure F.12).

The wide application of FET in digital radiography (DR) is due to their good performance in circuits and systems requiring high impedance.

**Filament circuit**  
(Diagnostic Radiology) This is the electrical circuit which supplies and controls the filament current \( I_f \), which determines the temperature of the cathode and, hence, the anode current \( I_a \) during the exposure. The filament circuit is diagrammatically presented as part of the high voltage generator and high frequency generator electrical circuits – see the eponymous articles.

**Abbreviation:** RT = Radiotherapy

**Related Articles:** Filament circuit, Beam weight

![FIGURE F.11 Symbols of field-effect transistors.](image-url)
The filament current \( I_f \) is supplied to the cathode through the filament transformer (FT) – a step-down transformer with ratio between 1:10 and 1:20. It generates a filament current on the order of 3–5 Amperes (A) through the cathode wire. The voltage of the secondary side of the FT is relatively low (around 10–20 V), but it is connected to the cathode which is at very high negative potential (kV). Due to this reason, special insulation is necessary between the primary and secondary coil of the FT. One way to achieve such insulation is by winding the primary coil over the core of the FT, placing over it a porcelain cylinder and winding the secondary coil over this cylinder. Additionally, the FT is placed together with the high voltage transformer (which generates the accelerating anode voltage, \( U_a, \) kV) in the high voltage box with isolating oil.

The current through the cathode filament (filament current \( I_f \)) depends on the maximum permissible power for selected focal spot \( P_{\text{max}} \) and the filament circuit assures anode current \( I_a < P_{\text{max}}/U_a \) for each selection of focal spot and anode voltage. In classical high voltage generators, a variable resistor is used to select the set the x-ray tube anode current \( I_a \), mA), by changing the voltage over the primary coil of the FT. In contemporary high frequency generators, the filament current is controlled by controlling the frequency of the current passing through the FT.

Normally the filament current keeps the cathode constantly preheated (with approximately 2 A filament current) in order to keep short the preparatory period (before the exposure), i.e. short time for initiating the flow of the anode current. The time for rapid change of the anode current depends on the thermal time-constant of cathode wire and could be on the order of 50–200ms. In case of grid-controlled tube (by the Wehnelt electrode), the filament circuit contains a subcircuit for supplying power to this additional electrode.

Related Articles: Cathode, Filament, Filament current, Filament heating, High voltage generator, High frequency generator

Filament current

(Diagnostic Radiology) This is the current which flows through the x-ray tube cathode filament and heats it. This current \( I_f \) is usually several Amperes and heats up the cathode to above 2000°C, regulating \( I_f \) through the filament voltage \( U_f \), the FT output voltage) and affects the number of thermal electrons created in the filament, hence the tube (anode) current \( I_a \).

The filament current is limited by the maximum permissible temperature of the filament coil. Thus the limiting factor for \( I_f \) at low kVp is the temperature of the filament (see the articles on Space-charge effect and Tube load), while the limiting factor for \( I_a \) at high kVp is the temperature of the anode. Saturation of the tube current \( I_a \) for anode voltages below 50kV is often called ‘space charge limited’ operation of the x-ray tube. Note, in Figure F.13, the limitation of the anode current \( I_a \) at 50 and 60kV due to the fact that all thermal electrons (created by the filament) have been extracted by the accelerating voltage \( U_a \).

The maximum filament current is limited to a set value. Its adjustment is made with different kV mA\(^{-1} \) settings using the so-called, three-point technique to properly select the necessary resistors in the filament circuit (for: low kV at high mA, high kV at low mA, low kV at low mA).

Often the filament emission chart is combined with another chart showing the relation between the filament voltage, \( U_f \), and filament current, \( I_f \). This is a linear relationship (often \( U_f \) change from 2 to 8 V creates \( I_f \) from 4 to 6 A).

For more details, see the article about Filament circuit.

Related Articles: Cathode, Filament circuit, Filament heating

Filament heating

(Diagnostic Radiology) The variation of the anode current (tube current) – the thermal electrons flying from cathode to anode – is achieved by changing the temperature of the cathode, which in turn is achieved by changing the filament current, \( I_f \) (Figure F.14).

![Filament emission curves (Ia as a function of If with Ua as a parameter)](image1)

**FIGURE F.13** Filament emission curves \( (I_a \text{ as a function of } I_f \text{ with } U_a \text{ as a parameter}) \).

![Emission current (density of electron flow) as a function of filament temperature](image2)

**FIGURE F.14** Emission current (density of electron flow) as a function of filament temperature.
The density of the thermal emission current is described by the Richardson equation:

\[ J = J_0 T^2 e^{-w/kT}, \]

where

- \( J_0 \) is the density of the emission current
- \( T \) is the temperature of the emitter (in K)
- \( k \) and \( w \) are constants (\( k \) = Boltzmann constant, \( w \) = work function, for tungsten = 4.5 eV)
- \( A_0 \) is the constant depending of the material of the emitter (for tungsten = 60 A cm\(^{-2}\) K\(^{-2}\))

The very high temperature of the filament leads to some evaporation of the tungsten wire. This evaporation leads to shortening of the life of the cathode (thinning it). Normally, the cathode life at this temperature is not more than 1000 working hours. Due to this reason, the cathode is heated to this high temperature for limited time only (during the x-ray exposure).

To heat the cathode from room temperature to 2700 K takes time; so, in order to keep the heating time short, the cathode stays always preheated at temperature around 1500 K. The preheating is achieved by applying a constant stand-by filament current through the cathode (less than 1 A). This way, the time to heat up the filament from the preheating to the requested temperature is much shorter (less than a second). When performing radiography, the operating x-ray switch normally has two phases (two steps button). The first stage of the button pressing (known as tube preparation, or prep stage) is associated with heating the cathode filament to the necessary temperature and rising the rate of anode rotation (in the case of a rotational anode x-ray tube). The second stage of the button pressing applies the high voltage and produces the exposure.

Figure F.15 shows the cathode of the x-ray tube during fluoroscopic operation (all filtration has been intentionally removed). The heated filament (the small glowing wire on the right) is seen in its focusing cup. Just opposite it (on the left) is the glowing hot rotating anode.

Being opposite to the very hot anode (at some 2.5 cm distance), the cathode filament is additionally indirectly heated from it (Figure F.15). This can affect the production of the thermal electrons and effectively create a positive feedback, which could destroy the x-ray tube (the very hot anode heats the cathode, thus increasing the cathode temperature; this increases the anode current, which additionally heats the anode, etc.). A special regulatory circuit with sensor compensates for the influence of the hot anode over the cathode temperature (this is part of the filament circuit).

The filament is heated with current, operating at frequency derived from the main frequency. The same frequency is applied to the anode voltage. This may cause a resonance effect and, thus, increased ripple. To prevent this phenomenon, modern x-ray equipments use an inverter in the filament circuit (which operates at higher frequency). In this case, pulse-frequency modulation can be used to give very precise control of the filament current.

**Related Articles:** Cathode, Filament circuit, Filament current, Tube current, Tube kilovoltage

**Filament (of an x-ray tube)**

(Diagnostic Radiology) The heated tungsten wire of the x-ray tube cathode produces thermal electrons. The electrical resistance of the cathode filament is relatively high and changes from approximately, 0.1–0.3 \( \Omega \) when cold, to 2–6 \( \Omega \) when heated above 2000 K (ohmic heating). For more information, see the article on Cathode.

**Related Articles:** Cathode, Filament current, Filament heating, Focal spot

**Filament resistor**

(Diagnostic Radiology) The filament circuitry of the x-ray generator includes a number of special resistors which effectively control the filament current through the cathode wire, hence the temperature of the cathode, and thus the anode current. Most of these resistors are connected in series with the variable resistor used to select the anode current (mA selector) and with the FT. The variation of resistance leads to change of the input voltage of the FT, hence its output voltage and filament current (see the articles on High voltage generator and Filament circuit).

There are sets of resistors which limit the filament current to a set value (depending on the x-ray tube type). These resistors are adjusted by the service engineer as per different kV mA\(^{-1}\) settings (for: low kV at high mA, high kV at low mA, low kV at low mA).

Another set of resistors are associated with the selection of the focal spot (i.e. selection of fine or broad filament wire). These resistors limit the maximal current which can pass through the respective wires.

**Related Articles:** Cathode, High voltage generator, Filament current, Filament heating, Filament circuit, Focal spot

**Fill-in factor**

(Diagnostic Radiology) See Detector fill factor

**Filling factor**

(Magnetic Resonance) The filling factor is a measure of the interaction of an RF coil with the object being imaged in that coil. It is defined as the ratio of the total magnetic energy in the transverse component throughout the volume imaged to the total magnetic energy throughout space for a given coil. The energy stored per unit volume in a magnetic field \( B \) can be shown to be

\[
\text{Energy per unit volume} = \frac{B^2}{2\mu_0}
\]
The energy stored in the rotating transverse component $B_1$ (i.e. the detectable component) in MRI is

\[
\text{Energy} = \frac{\int B_1^2 dV}{2\mu_0}
\]

The filling factor is then

\[
\text{Filling factor} = \frac{\int B_1^2 dV}{\text{Total energy}}
\]

For a simple coil, the total energy stored is

\[
\text{Total energy} = \frac{1}{2}LI^2
\]

where

- $I$ is the current in the coil
- $L$ is the coil inductance

Filling factor clearly influences signal to noise ratio (SNR). Higher filling factors can be achieved by closer conformity of a given coil to the anatomy of interest (e.g. array coils or flexible surface coils).


**Film badge**

*Radiation Protection* Film badges (Figure F.16) are used for personal dosimetry of the staff exposed to radiation, such as those employed in x-rays and nuclear medicine laboratories as well as in radiotherapy units.

The film is sensitive to x-rays and gamma radiation but also to light – therefore, the film must be kept in a light-tight holder. The film consists of the emulsion composed of crystals of silver bromide AgBr (Ag⁺ and Br⁻) which is stuck with a gelatine to a polyester base. When the emulsion is irradiated with X or gamma photons, free electrons are created by the photoelectric effect and Compton scattering process. These free electrons remove the electrons from bromide and neutral atoms of Br are absorbed by gelatine. The electrons passing through crystal neutralize silver ions. As a result of the irradiation, a latent image is produced in the crystal. To see this image, it is necessary to develop the film using a procedure depending on the type of film. After the development, fixation and hardening the latent image is observed in the form of dark silver grain speck on the film which is called blackening. Its quantity is measured by its optical density $D$ defined as

\[
D = \log\left(\frac{I_0}{I}\right)
\]

where

- $I_0$ is the incident intensity of visible light
- $I$ is the transmitted intensity by the developed film

In Figure F.17, the dependence of the density $D$ on the exposure (log $E$) is shown. It is linear in the region corresponding to exposure values between $E_1$ and $E_2$. The film may be used to determine the dose after calibration using only the linear region. To produce in air an optical density of one, the dose of about $2 \times 10^{-4}$ Gy is required for a typical film.

The film badge holder usually contains a set of filters (e.g. of cadmium to capture neutrons in [n, γ] reaction). The example is shown in Figure F.18.

**Related Articles:** Dosimeter, Integrating dosimeter
Film base

(Diagnostic Radiology) The base of a typical radiographic film is made of a clear polyester material about 150 μm thick as illustrated in Figure F.19. It provides the physical support for the other film components and does not participate in the image-forming process. In some films, the base contains a light blue dye to give the image a more pleasing appearance when illuminated on a view box.

Film blackening

(Diagnostic Radiology) Film blackening is the result of exposure to a film that is then chemically processed. For radiographic films, the more precise term for film blackening is optical density.

Film digitisers

(Diagnostic Radiology) Film digitisers are instruments that convert images recorded on film into digital images. Conventional scanners used with computers are one type of image digitiser but usually not designed to scan transparent films like radiographs. Specific features of digitisers designed for radiographic film include a large area of coverage (for chest films, etc.) and a rear illumination source for scanning transparencies (Figure F.20).

The most important parameter which characterizes a film digitiser for radiographic purposes is the digital image pixel size, commonly quantified in dot per inch (DPI). This is related to the requested spatial resolution and the pixel depth, set in number of bits per pixel, which is related to the dynamic range of the grey levels of the pixel.

Film dosimetry

(Diagnostic Radiology) See Film badge

Film emulsion

(Diagnostic Radiology) The emulsion is the active component in which the image is formed and consists of many small silver halide crystals suspended in gelatine. The gelatine supports, separates and protects the crystals. The typical emulsion is approximately 10 μm thick. The emulsion is coated onto the film base which provides a rigid support. Radiographic film can be made by double layer emulsion in order to increase the sensitivity or by single layer emulsion in order to increase the spatial resolution.

Film fog

(Diagnostic Radiology) In conventional radiology, fog is any undesirable density or darkness that appears in a film that is not a result of the exposure forming the image. Sources of radiographic film fog include:

- Exposure to other sources of x-ray or gamma radiation
- Accidental exposure to light (defective cassettes, light leaks into darkroom, etc.)
- Overdevelopment (called chemical fog) because of high developer temperature, excessive development time, incorrect or contaminated chemistry
- Storage of unexposed film for long times and at high temperatures

Fog adds to the density in the ‘toe’ region of a film characteristic curve. That is the region of the curve relating to very low or no applied exposure. The optical density value of unexposed film is due to the intrinsic density of the film base plus any other unexpected factors giving rise to film fog. Therefore, the optical density of an area in a film that has not been exposed is designated as the ‘base + fog density’ (the optical density of such area is normally below 0.2D).

Related Article: Characteristic curve

Film holder

(Diagnostic Radiology) Film holder (also known as cassette) is a thin flat light-tight container for the x-ray film (and the respective phosphor-intensifying screens). The cassette should keep the film out of the reach of any light, put the film and screens in very close contact and allow the x-rays to pass through and expose the film. The front side of the cassette has minimal x-ray absorption. One intensifying screen is mounted inside this part of the cassette. The back side of the cassette has to be only sturdy, as x-ray absorption is not more important. This side can also include a thin lead sheet to prevent backscatter from the patient table to expose the film from the back. Another intensifying screen is mounted inside this part of the cassette. Usually, this screen is thicker than the entrance screen. The film holder (cassette) is kept locked with special latches; these are different in the case of hand development or automatic development of the film. Computed radiography systems use similar-sized holders (but with different inner components) to hold the storage phosphor.

Related Articles: Cassette size, Cassette carriage, Screen film contact

Film processing
(Diagnostic Radiology) Film processing is the procedure of transforming the latent film image to a visible one. Classical x-ray radiography relied very much on film processing. This chemical process included a number of specific phases: film development, followed by fixing, washing and drying. All these are described in detail in their specific articles.

In short, the photographic film emulsion contains silver bromide (AgBr) and free silver ions moving within its cubic lattice. Each crystal also includes specific impurities (such as sulphur), which form crystal defects – electron traps. When a light (or x-ray) photon excites a bromine atom, it loses an electron. These free electrons are trapped into the crystal defects. The positive free silver ions are attracted into these negative defects, where they are neutralised and become Ag atoms (sensitised grains). The combination of areas in the film with different number of sensitised grains forms a latent image.

During the process of film processing, the emulsion is first developed – a chemical process during which the sensitised grains are stabilised (the exposed AgBr crystals reduced to stable Ag atoms). During the next chemical process of film fixing, the remaining unsensitised grains (which had not been exposed to light photons) are removed and washed out. The film is then dried and ready to be seen on a viewing box. The final visible image contains areas with variable darkness (depending on the concentration of opaque Ag atoms), which is proportional to the intensity/energy of the light or x-ray photons.

**Related Articles:** Film emulsion, Silver bromide, Latent image, Developer, Fixer, Fixing agent, Acetic acid in film processing, Accelerators in film development, Thiosulphate in film processing, Washing in film processing, Dark room, Automatic film processor, Film transport, Underdevelopment


Film screen contact
(Diagnostic Radiology) Film screen contact refers to the closeness of contact between the film and the surface of the intensifying screens in a radiographic cassette. If there is space (not good contact) between a film and screen, it produces blurring and reduces visibility of detail. There are established tests for film screen contact within quality control procedures (Figure F.21).

**Image blurring poor film-screen contact**

**FIGURE F.21** Image blur due to poor film-screen contact. (Graphs courtesy of Sprawls Foundation, www.sprawls.org)

**Film transport**
(Diagnostic Radiology) The film transport is the mechanism within a film processor that moves or transports film through the developer, fixer and wash tanks. It consists of a series of rollers and guides as shown in Figure F.22.

The film transport can be easily removed from the processor for cleaning as shown in Figure F.23.

**Film type**
(Diagnostic Radiology) A film is classified by its characteristics. For radiography, there are different types of films designed for specific clinical applications. The two major characteristics that are associated with film type are exposure sensitivity (speed) and contrast.

**Related Articles:** ASA, Speed of film

**Filter**
(Nuclear Medicine) This article refers either to the filtering applied in image processing or physical filtering of particles.

**Image Processing:** Software filters in medical images are used to increase the diagnostic value of the collected images. Example of filters that are used are smoothing filter (to reduce noise) or edge enhancement filter (to increase contrast in regions with a high signal gradient).

**Physical Filtering:** The process of physical filtering involves passing a quantity of water or gas through a filter. Examples of physical filtration in nuclear medicine are the filtration of colloid particles (see the article Filtration) and the filtration of 11CO-gas to prevent 11CO2-breakthrough in 11C production using a cyclotron.

**Related Article:** Filtration


**Filter, compensating**
(Diagnostic Radiology) These are metal (most often aluminium) absorption filters used in radiography to obtain images with more uniform contrast. They are especially useful in anatomical regions with significant absorption variation (chest radiography, skull radiography, etc.). The filter used in chest imaging is Trough filter with less absorption in the middle (where the image of the spine will be) and more attenuation at the edges (where the edges of the lungs will be imaged). Such filters are also useful in angiography (and specially digital
absorption angiography (DSA). The filters are also known as wedge filters. For more information, see the article on Beam restrictor.

**Related Articles:** Attenuation, Beam restrictor, Diaphragm, Collimator

**Filtered back projection**

*(Nuclear Medicine)* Filtered back projection is a reconstruction method used to reconstruct 2 dimensional (2D) images into a 3 dimensional (3D) volume.

In emission tomography, imaging projections are acquired at discrete angles $\alpha_i$. These projections are a 2D representation of the source distribution. Using the image reconstruction technique called back projection, it is possible to get a 3D source distribution. If one accepts a few simplifications, such as the absence of scattered events, then one row in each projection can be considered a 1D representation on the source distribution along a line. The 1D projections from different angles originating from the same position along the trans-axis can be collected in a sinogram. The first row in a sinogram is the 1D projection from angle $\alpha_i$ and the second is $\alpha_{i+1}$ and so forth until all the projection angles are represented. The number of projections acquired determines the number of rows in the sinogram.

In simple back projection, each projection is extracted from the sinogram and placed in all rows in an *image matrix* (the number of columns is equal to the number of columns in the projection and the number of rows can be arbitrarily chosen, but typically they are the same). This matrix is rotated according to the corresponding projection angle. This procedure is repeated and the rotated image matrices are summed together to produce a reconstructed 3D image. Images reconstructed using simple back projection do not have an adequate spatial resolution.

**Filtered Back Projection:** Because of the low image spatial resolution achieved with simple back projection, a number of filters can be applied to the individual projections to enhance the resolution. Before filtering, the projection data are transformed using the Fourier transform. Using the Fourier transform, spatially varying data can be expressed as a series of sine and cosine functions. In transformed data, the functions with high frequencies represent sharp edges and low frequencies large image features. A filter that amplifies high-frequency components relative to low-frequency components will increase the spatial resolution but at the same time lower the signal to noise ratio. The projection data are then transformed back to the spatial domain using an inverse Fourier transform.

The filtered projections are then used to reconstruct a 3D image with simple back projection.

**Related Article:** Pile up effect


**Filtration, inherent**

*(Diagnostic Radiology)* See *Inherent filtration*

**Filtration (of colloidal particles)**

*(Nuclear Medicine)* Filtration is a method for physical quality control of colloidal particles, by the use of membrane (ultra) filtration or gel filtration. In both cases, the measurable quantity is the size of the particle, which is obtained by measurement of either the activity of the sample or the number of particles as a function of their sizes. In general, the activity of a particle is proportional to the volume or surface of the particle. Thus, larger particles have normally considerably higher activity than smaller particles. It should be remembered that the activity distribution obtained does not give the size distribution.

Practically, small samples of the colloids are passed through polycarbonate filters (Nuclepore© or Millipore®). The pore sizes of the filters are available between 15 and 1000 nm. The filter is rinsed after the passage of the colloid (0.01–0.25 mL sample) with 1–4 mL distilled water. The filter and the filtrate are then measured for the activity. The activity in the filtrate is given as percentage of the total activity.

Physical control by the use of filtration of a new radiopharmaceutical, primary regarding particle size, concentration and stability may be important prior to functional investigations, for example of the reticuloendothelial system (RES). The characteristic of the particles is very important for the uptake rate and the distribution in the tissue. Different techniques for particle characterisation have been published by Bergqvist et al.

Membrane filtration is also the most common method for sterilisation of radiopharmaceuticals, using a Millipore filter for removal of various organisms. A common membrane filter size is 0.45 µm,
but a pore size of 0.22 μm is used for the sterilisation of products of the blood and preparations containing smaller microorganisms.

**Related Articles:** Biological purity, Quality control, Tc-99m-labelled microcolloids, Tc-99m-labelled nanocolloids, Tc-99m-albumin microcolloid, Tc-99m-albumin microspheres, Tc-99m-albumin nanocolloid, Tc-99m-rhenium sulphide colloid


**Filtration rate**

(*Nuclear Medicine*) Filtration rate is the rate of fluid filtered through the kidneys per unit time. It is also known as glomerular filtration rate (GFR). A deviant filtration rate can be a sign of a pathological condition and the parameter can be estimated using Tc-99m-DTPA and a SPECT camera.

**Abbreviation:** SPECT = Single photon emission computed tomography.

**Related Articles:** Tc-99m-DTPA, Single photon emission computed tomography (SPECT)

**Filtration, total**

(*Diagnostic Radiology*) The total filtration of an x-ray beam aims to reduce the unnecessary, low-energy x-ray photons. It is produced by a combination of two filter components – inherent and added. The inherent filtration consists of existing components of the x-ray system through which the x-ray beam passes (tube window, oil in tube housing, beam locator mirror, etc.). The added filtration of the x-ray beam comprises of sheets of metal (most often aluminium) that are added to provide the required total filtration.

Total filtration is monitored by international and national regulations (e.g. all x-ray equipment capable of producing energies above 100 kV must have total filtration at least 2.5 mm aluminium equivalent.

**Finger ring dosimeter**

(*Radiation Protection*) Finger ring dosimeters (Figures F.24 and F.25) are made from the thermoluminescent material (e.g. LiF:Mg:Ti).

They are used as an integrating dosimeter to measure the radiation dose to the fingers of staff, mainly those in interventional radiology and staff working in nuclear medicine laboratories.

Typical range of response: 0.1 mSv–1 Sv for the radiation of energy of 15 keV–3 MeV. The dosimeters can be used to integrate doses over periods of time from 1 to 3 months.

**Abbreviations:** TL = Thermoluminescence and TLD = Thermoluminescent dosimeter.

**Related Articles:** Integrating dosimeter, Thermoluminescent dosimeter (TLD)


**FISP (fast imaging with steady state precession)**

(*Magnetic Resonance*) See Fast imaging with steady state precession (FISP)

**Fixation**

(*Radiotherapy*) See Immobilisation

**Fixed aperture beam restrictors**

(*Diagnostic Radiology*) See Beam restrictor

**Fixer**

(*Diagnostic Radiology*) After leaving the developer, the film is transported into a second tank, which contains the fixer solution. The fixer is a mixture of several chemicals that perform the following functions:

**Neutraliser:** When a film is removed from the developer solution, the development continues because the solution was soaked up by the emulsion. It is necessary to stop this action to prevent overdevelopment and fogging of the film. Acetic acid is in the fixer solution for this purpose.

**Clearing:** The fixer solution also clears the undeveloped silver halide grains from the film. Ammonium or sodium thiosulphate is used for this purpose. The unexposed grains leave the film and dissolve in the fixer solution. The silver that accumulates in the fixer during the clearing activity can be recovered; the usual method is to electroplate it onto a metallic surface within the silver recovery unit.

**Preservative:** Sodium sulphite is used in the fixer as a preservative.
**Hardener:** Aluminium chloride is typically used as a hardener. Its primary function is to shrink and harden the emulsion.

**Related Article:** Film processing

**Fixing agent**

*(Diagnostic Radiology)* After leaving the developer, the film is transported into a second tank, which contains the fixing agent solution. The fixing agent is a mixture of several chemicals that perform the following functions (see the article on *Fixer*).

**Related Article:** Fixer

**FLAIR (fluid attenuated inversion recovery)**

(Magnetic Resonance) See *Fluid attenuated inversion recovery (FLAIR)*

**FLASH (fast low angle shot)**

(Magnetic Resonance) See *Fast low angle shot (FLASH)*

**Flat field image**

*(Diagnostic Radiology)* A flat field image is an x-ray image taken with no object in the beam and is thus, a low contrast image, usually appearing black or dark grey. It can be used during equipment evaluation and quality control to calculate the quantum and system noise of the x-ray imaging equipment. A flat field image must be taken to determine the noise power spectrum (NPS) and thus the noise equivalent quanta (NEQ) and detective quantum efficiency (DQE) of the system.

**Related Articles:** Detective quantum efficiency (DQE), Noise equivalent quanta (NEQ), Noise power spectrum (NPS), Signal to noise ratio (SNR)

**Flat panel array**

*(Diagnostic Radiology)* See *Flat panel detector*

**Flat panel detector**

*(Diagnostic Radiology)* Flat panel detectors are used as digital radiographic (x-ray) detectors and can be divided into two main subgroups: direct and indirect detectors, Figure F.26. They are usually of slim design to allow them to be used in conjunction with the traditional Bucky tables and have dimensions of at least $35 \times 43$ cm$^2$ for general radiography.

They are typically constructed from a large area, active matrix array which is electronically coupled to a solid-state detector, either a photoconductor or a phosphor screen. Direct detectors convert the incident x-rays directly to electrical charge by the use of a photoconductor, such as amorphous selenium. Selenium ($Z = 34$), usually used in direct detectors, has a higher atomic number than silicon ($Z = 14$), and, therefore, its attenuation coefficient is higher. In contrast, indirect detectors convert the incident x-rays to electrical charge by a two-stage process. Firstly, a scintillation phosphor screen, such as caesium iodide doped with thallium (or gadolinium oxide sulphide), converts the incident x-rays into visible light wavelength photons and then a photodetector, such as an amorphous silicon photodiode array, converts the visible light photon to electrical charge. Some radiographic systems including charge-coupled devices (CCD) are sometimes also termed indirect flat panel detectors as they use a large-area scintillation phosphor screen to convert the incident x-rays to visible light photons; but, as the released light photons are then typically focused onto a smaller CCD integrated chip, these are not strictly considered to be flat-panel, large-area devices and do not use active matrix arrays for the readout process.

Excluding CCD-based devices, the readout technology used for both direct and indirect detectors is based upon technology designed for LCD monitors and thin film technology. To display an image on an LCD monitor, a large two-dimensional array of thin film transistors (TFTs) arranged on a glass substrate is used to charge and consequently illuminate individual pixels to achieve the desired intensity. To address each pixel element individually, instead of wiring each element separately, they are connected together by a series of horizontal and vertical lines, and thus, by using the correct logic, each individual pixel can be accessed and addressed. This type of array is called an active matrix array. The active matrix array used in flat panel detectors is an integrated circuit formed out of a large number of photodetector elements connected to TFTs. It can be produced as a large area matrix (currently available $43 \times 43$ cm$^2$). The active matrix array for a flat panel detector works in an opposite fashion to an LCD monitor. The signal for each detected incident x-ray photon is stored as charge in each pixel element, and the array is then used to read out this charge from sequential pixel elements, row-by-row, in an active matrix readout.

*Figure F.27* shows a typical array used in a medical imaging flat panel detector. Within the array, each pixel consists of the switching element and an element to detect incoming photons and store them as charge. The image read-out process is controlled by altering the voltage applied across the switching element. Firstly, to allow each pixel to detect a signal during exposure, the voltage across each switching element is set to an ionisation or ‘off’ state. The signal is then read out by changing the switching voltage row-by-row to the conducting or ‘on’ state which allows the charge stored in each pixel to be drained by the charge collector electrode and passed to the multiplexer. The voltage change is controlled by the gate line driver. As the read-out process is controlled by the external circuitry, each row of pixels requires a separate control line driver to alter the switching voltage, and each column its own amplifier. This process is called the active matrix read-out.

**Abbreviations:** TFT = Thin film transistor and CCD = Charged coupled device.

**Related Articles:** Direct x-ray detection, Amorphous selenium, Active matrix array, Thin film technology, LCD monitors, Silicon diode detector

**Flattening filter** *(Radiotherapy)* The flattening filter is located within the head of a linear accelerator (just below the primary collimator) and is used to produce a ‘flat’ beam profile for x-ray photon radiotherapy treatment. The flattening filter is typically made from aluminium, and different flattening filters are used for different beam energies and can be replaced in the beam path by scattering foils whenever an electron beam has to be produced.

The beam produced from bombarding the target with electrons results in a very strongly forward peaked beam. By placing a flattening filter in the beam path, this peaked profile is modified to a more usable flat profile as illustrated in Figure F.28. It is important that the beam steering and angle of impact on the flattening filter is correct; otherwise, a sloping profile will result as illustrated in Figure F.29.

Therefore, it is clear that the position of the flattening filter with respect to the beam central axis is critical, and any movement of either will produce a significantly different profile which may be asymmetrical and no longer flat. It is also important that the energy of the beam is monitored as changes can cause the beam to be either over- or under-flattened. The beam uniformity is measured within the head of a linac by the monitor chamber.

Those linacs that operate with more than one x-ray energy will also have more than one flattening filter, and there will be an interlock system in place to avoid the incorrect filter being used.

**Related Articles:** Optimal incident beam profiles, Linear accelerator, Treatment head, Beam flatness, Beam symmetry, Monitor chamber


**Flicker** *(General)* Flicker refers to the visual ‘flickering’ effect seen on video displays. It is particularly evident on computer displays based on the CRT.

The display screen of a CRT device is a glass screen coated on the inside with a phosphorescent material. An electron gun provides a focused beam of electrons which strike the phosphor at a high velocity and cause it to give off phosphorescent light. The lifetime of the light emission is called the persistence, and this varies depending on the phosphor material. If the persistence time is short, the image will fade and will have to be refreshed. Flicker occurs if the refresh rate is low. A refresh rate of 60Hz, for example may exhibit flicker, whereas a rate of 70Hz or higher will usually result in flicker-free viewing.

**Flip angle** *(Magnetic Resonance)* The flip angle (α) describes the net rotation of the bulk magnetisation vector by an RF pulse. At resonance, it is determined by the time integral of $B_1$: 

![Diagram](image-url)
of the static magnetic field (Figure F.30). The remaining longitudinal magnetisation, $M_L$, and transverse magnetisation in the $x$–$y$ plane.

$$
\alpha = \gamma \int B_I(t) dt
$$

For an excitation pulse, it is given with respect to the direction of the static magnetic field (Figure F.30). The remaining longitudinal component influences the $T_1$-weighting in steady state. For a refocusing pulse, the flip angle influences the echo amplitude.

**Related Article:** $B_I$

**Float value**

(General) In computing, numbers are stored in a representation that depends on the storage capacity of the system and the required accuracy of the number. A number is generally encoded in a string of digits, which can be decoded if the computing system has information on the coding representation. Float value, or floating point representation, is one such method that allows storage of high precision values over a wide range.

For example, floating point would be a better representation to store the number ($\rho$), an irrational number whose decimal representation hence never ends, than the integer representation.

The name ‘float’ is given because the representation allows the decimal point to ‘float’ anywhere relative to the significant digits. The stored number would consist of a signed digit of a given length (the significand), and a signed integer exponent, which modifies the magnitude of the number.

For example, in decimal notation, the number 4562.907 is represented as the significand 4.562907, together with an exponent of 3. Floating point can be used with different bases other than 10 (decimal).

Equation F.1 is a symbolic representation of floating point representation in decimal notation:

$$
\text{Value} = \text{significand} \times 10^{\text{exponent}} \quad (F.1)
$$

E.g. 4562.907 = 4.562907 × 10^3

Because of the flexible positioning of the decimal point, floating point representation can achieve a much wider range of values for the same storage requirements. However although float values are generally of high precision, they cannot faithfully reproduce a real irrational number, and this limiting precision should always be considered in calculations and storage.

**Flood field**

(Nuclear Medicine) Flood field images are acquired when the detector is irradiated by a parallel beam of photons with a perpendicular angle of incidence covering the entire detector surface.

Flood field images are used to measure image non-uniformity in scintillation cameras. Uniformity is the camera’s ability to depict a uniformly distributed field of photons. Camera uniformity is controlled on a regular basis as a part of the camera quality control process. If the measurements show signs of non-uniformity, mathematical corrections should be applied. The uniformity can be measured in two ways: the intrinsic and extrinsic uniformity. *Intrinsic flood field* images are acquired without the use of a collimator. The source depicted is usually a point source placed far away from the detector (typically four to five times the diameter). Such experimental set up will guarantee a near uniform radiation of the detector.

Extrinsic flood field imaging is performed using a collimator and a uniform flood source placed close to the collimator as possible. The area of the flood source must cover the detector area.

The information acquired in the uniformity measurements are then used to update the uniformity correction used for patient examinations.

Related Article: Scintillation camera


**Flood field image**

(Nuclear Medicine) Flood field image refers to an image used to quantify the uniformity of the detector response. According to National Electrical Manufacturers Association (NEMA) standards, a (intrinsic) flood field image for a scintillation camera should be acquired using $99m$Tc in a source holder (to prevent side and backscatter) at a distance greater than five times the detector diameters. The detector surface should not be collimated. The count rate should not exceed 20,000 counts per second for small FOV detector. For high count rates, the detector surface can be shielded by a lead mask with homogenous
thickness. Image acquisition should be continuous until the central element has registered 10,000 events per pixel. Before evaluation, the image should be convolved with a smoothing filter. To read more about evaluation of the flood field image, please see the article Uniformity.

**Related Articles:** Scintillation camera, Uniformity

**Further Reading:** National Electrical Manufacturers Association. 2007. NEMA Standards Publication NU 1. National Electrical Manufacturers Association, Rosslyn, VA, pp. 9–10.

### Flow compensation

(Magnetic Resonance) During an MR sequence, the object imaged is expected to be at rest. If this is not the case, and the object moves during and/or between data acquisition, artefacts such as mis-positioning and blurring may occur in the reconstructed images.

A moving spin will accumulate a phase offset when exposed to a gradient. This phase is different from the phase of a static spin (see Phase contrast and Velocity mapping). The offset phase angle results in mis-positioning and blurring of the object in the Fourier reconstruction (Figure F.31).

Flow compensation can be performed by introducing additional gradient lobes prior to the echo readout. The aim of these lobes is to cancel out the phase offset induced by motion.

Flow compensation is normally used when large vessels are in proximity to the desired object in the image. A consequence of flow compensation is longer TE, as more gradients need to be introduced prior the echo read-out.

Compensation both for velocity and for higher order movement such as acceleration and jerk can be done (see the article Acceleration compensation).

**Related Articles:** Acceleration compensation, Phase contrast, Velocity mapping

### Flow effects

(Magnetic Resonance) Many types of flow effects (effects on signal in MR images) are known. Among the most prominent are as follows:

1. **Wash-in/wash-out phenomena**
   Sequences using two or more spatially selective RF pulses, such as the SE and IR sequences, are sensitive to the transport of spins into and out of the excited slice. If the flow is coherent over at least a number of voxels (macroscopic flow) and has components perpendicular to the imaging slice, the signal will decrease due to the outflow of spins during the time between RF pulses. On the other hand, if the repetition time is not too long, the inflow of ‘fresh’, non-saturated spins during the repetition time will increase the magnetisation compared with the static, saturated, spins and thus give an increase in signal. These competing mechanisms make the modulus signal behaviour versus flow velocity biphasic in such sequences. If only one RF pulse is used per repetition time, as in gradient echo sequences, the signal decrease due to wash-out will not be observed. (See also Inflow effect and Flow void.)

2. **Phase effects due to macroscopic flow**
   The movement of spins affects the voxel magnetisation phase angle in quite a different way than the corresponding amplitude. The underlying mechanism has been known for some time and involves motion of spins along a magnetic field gradient. The phase shift caused by linear movement during a time \( t \) along a constant magnetic field gradient \( G \) in the absence of RF pulses is proportional to the velocity. Movements of higher order, i.e. accelerations, also affect the phase angle, but if these effects can often be disregarded, the voxel phase angle is a sensitive and reasonably good measure of linear flow velocities independent of relaxation parameters. (See also Phase mapping.)

3. **Pulsation effects**
   Contrary to steady flow, which will not interfere with the spatial phase encoding, a variable flow resulting in a varying phase shift during the measurement generally leads to artefacts or mis-positioning along the phase encoding direction. If the flow is pulsative and no physiological triggering or gating is applied, typical artefacts appear in regular patterns. Hence, studies of pulsative flow as well as examinations where a suppression of such artefacts is essential should involve a triggering/gating procedure.

4. **Phase dispersion effects**
   If spins within a voxel have a distribution of velocities along a certain gradient direction, a range of different phases results. This phase dispersion in the voxel reduces the net magnetisation vector amplitude but the net phase angle still reflects the net flow. If the dephasing effects are strong, however, and possibly combined with outflow effects, the signal may be destroyed completely (see also Flow void).

5. **Attenuation due to microscopic flow**
   Similar to phase dispersion effects, incoherent intra-voxel motion (diffusion, perfusion) in a non-homogeneous magnetic field can result in detectable phase dispersion, and hence to a corresponding reduction in amplitude. These effects are studied with dedicated MR techniques (see also Diffusion imaging and Perfusion imaging).

**Related Articles:** Diffusion imaging, Flow void, Inflow effects, Perfusion imaging, Phase mapping, Velocity encoding, Flow quantification

### Flow encoding

(Magnetic Resonance) Flow encoding is also known as velocity encoding. For further information, see the article Velocity encoding (VENC).

**Related Article:** Velocity encoding (VENC)

### Flow imaging

(Magnetic Resonance) A large number of methods for in vivo quantification of blood flow (flow imaging) using MRI have been proposed during the last three decades. Among the first scientists to explore the field were Herfkens et al. (1981), who stated that the information in a conventional MRI was flow dependent; Grant and Back (1982), who showed wash-out, or flow void, effects in tubes, and Crooks et al. (1983), who published a signal versus velocity
curve obtained with a spin-echo sequence. Singer and Crooks (1983) as well as Wehrli (1985) proposed quantitative methods for velocity measurements in vessels, based upon wash-in/wash-out effects. Magnetic tagging methods, often known as bolus tracking methods have been proposed for use in MRI by, for example Shimizu (1986).

With respect to flow-induced phase effects on motion, the attenuation in modulus images due to loss of phase coherence within the voxel was used by Waluch and Bradley (1984), who pointed out the clinical possibilities of the so-called even echo rephasing effect.

Development along another line utilized motion-induced phase effects more directly. Here, Moran (1982) proposed a method for the creation of velocity images by the addition of flow-sensitive or flow-encoding gradients to a standard pulse sequence. The technique was rapidly developed by, among others, van Dijk (1984) and Nayler (1986), leading to the today well-established method known as velocity mapping or phase mapping technique.

As a result of the work outlined earlier, quantitative flow measurement sequences and evaluation tools have become available on standard MRI scanners, providing the possibility not only to measure velocity and flow, but also to evaluate important physiological parameters, such as shear stress and compliance.


Flow imaging (Ultrasound) In ultrasound, the term flow imaging covers a range of techniques to image and measure flow and flow parameters by what are generally regarded as Doppler methods.

The most commonly used techniques are as follows:

- Continuous wave (CW). Non-depth-specific ultrasound device that gives an audio output of the Doppler frequencies. A spectral analyser can be added to give a sonogram display of the time varying distribution of the Doppler frequencies (Figure F.32).
- Colour flow imaging (CFI) uses pulsed wave Doppler to produce a colour-coded map of Doppler shifts in an area of an image superimposed on B-mode image.
- Power Doppler. Similar to CFI, power Doppler uses the amplitude of the CFI process output to display the presence of Doppler shifts and variation of the amount of moving scatterers in the image.
- Pulsed wave (PW) spectral Doppler. PW systems permit detailed examination of the circulation at a particular depth along a beam. In duplex ultrasound systems, they can be used with the B-mode and CFI image to produce a sonogram from a specific sample volume identified in the image (Figure F.33).

Other techniques used include

- Time domain colour flow imaging, similar to CFI autocorrelation techniques.
- Colour M-mode displays, showing the time-varying distribution of velocity vectors along a specific scan line.
- Contrast agent wash-out and wash-in. The uptake of contrast agents and measurement of the intensity of contrast agent density have been used as a measure of perfusion and transit time.

Flow phantom (Ultrasound) A phantom using fluid flow to test the Doppler and colour flow characteristics of a Doppler ultrasound scanner or device. See Doppler phantom
Flow quantification

(Magnetic Resonance) Two different methods exist for flow quantification with MR.

Modulus-based techniques rely on the change in the signal magnitude which occurs when moving spins travels out of or into the imaging slice. With suitable models, a quantification of the flow is possible. For more information, see Inflow-effect and Flow void.

Phase-contrast techniques are more common and use the signal phase for flow quantification. These methods use a bipolar gradient shape in the pulse sequence to extract velocity information from the phase value of the moving spins. The most common phase technique is phase-contrast MRI.

Accurate quantification of flow with phase-contrast techniques requires the flow encoding direction to be properly aligned with respect to the flow direction. Additionally, in the case of phase-contrast MRI, the spatial resolution needs to be high in order to minimise the effects of intravoxel dephasing and partial-volume effects and to have a suitable number of voxels in the blood vessels.

For more information, see Flow imaging and Inflow effect.

Related Articles: Flow void, Inflow effect, Phase contrast


Flow-sensitive alternating inversion recovery (FAIR)

(Magnetic Resonance) FAIR is an arterial spin labelling (ASL) technique for non-invasive perfusion measurements, in which tissue magnetisation is inverted in both a slice-selective (ss) and a non-selective (ns) experiment, while the arterial magnetisation is inverted only in the non-selective experiment (Figure F.34). In the slice-selective experiment, a 180° RF pulse is applied to the imaging slice, and after a labelling delay TI, the image is acquired in the same region. In the non-selective experiment, the magnetisation of the imaging slice as well as of the inflowing blood is inverted and the image is acquired (after the same delay TI as in the slice-selective experiment). The subtraction of the ns-based image from the ss-based image gives a perfusion-weighted difference map. The FAIR labelling technique is optimal in cases where the vascular geometry is complex, for example in the heart, lungs and kidneys. Perfusion quantification using FAIR is accomplished by the following equation:

\[
\Delta M(TI) = 2\alpha M_0 f \lambda T I e^{-\lambda T I} T I
\]

where

- \(\Delta M\) is the perfusion-weighted magnetisation difference between the two experiments
- \(\alpha\) is the degree of inversion (for complete inversion \(\alpha = 1\))
- \(M_0\) is the tissue equilibrium magnetisation
- \(f\) is the tissue blood flow
- \(\lambda\) is the tissue-to-blood partition coefficient
- \(T_1\) is the \(T_1\) of tissue

Related Articles: Perfusion imaging, Arterial spin labelling, EPISTAR, PICORE, QUIPSS – QUIPSS II – Q2TIPS


Flow void

(Magnetic Resonance) The term flow void is used for a frequently observed signal loss phenomenon in MRI, caused by flowing material such as blood in vessels. Flow voids can have at least two technical explanations:

1. In pulse sequences, where at least two RF pulses are executed per repetition time interval (TR), a through-plane motion of material during sequence execution will have the effect that not all spins experience the desired pulse combination, thereby causing signal loss. For example, consider a spin-echo-sequence where blood flows in a vessel oriented perpendicular to the imaging plane (figure below). The blood experiencing the 90° pulse (hatched) will, during the time period between the 90° and the 180° pulse (TE/2), move out of the slice and the spin echo is formed only in the part of the blood vessel that experienced both pulses (crossed); thus, the signal is reduced in proportion to the amount of blood that left the slice during TE/2. This phenomenon is also commonly known as the wash-out effect.

   It should be noted, that in sequences requiring only one RF pulse per TR, such as a conventional Gradient Echo (GRE), the wash-out effect can normally be ignored.
2. When magnetic field gradients are applied, spins moving in the gradient direction exhibit a phase evolution that differs from that of static spins (see Velocity mapping). After execution of a specific gradient pulse waveform, the phase offset for spins in motion will depend on the specific gradient waveform as well as the velocity. Hence, the phases will disperse in a volume of interest (voxel) covering a distribution of velocities, and the net signal will be reduced. The magnitude of this dephasing effect may be substantial, and it depends upon imaging parameters, such as voxel size and gradient strengths, as well as upon motion direction and magnitude.

**Related Article:** Velocity mapping

**Fluence optimisation**

*(Radiotherapy)* Fluence optimisation is a technique used in radiotherapy treatment plan optimisation, particularly in external beam x-ray planning. For each treatment beam, an optimisation algorithm is used to generate the fluence profile needed to deliver the prescribed dose distribution using inverse planning. After the plan has been optimised, a second stage is often involved in which the constraints of the delivery system are modelled to produce a deliverable beam plan.

**Related Articles:** Interactive planning, Inverse radiotherapy planning, Simulated annealing algorithm

**Fluid attenuated inversion recovery (FLAIR)**

*(Magnetic Resonance)* The aim of the FLAIR sequence is to suppress fluid signal by an inversion recovery preparation at an adjusted inversion time (TI). The TI value in the IR sequence is chosen in press fluid signal by an inversion recovery preparation at an adjusted relaxation time and is given by $T_{1\text{ null}} = 0.69T_1$ for sequences with long repetition time $TR$. Since the $T_1$ relaxation time of fluid is large, a long inversion time on the order of 2000 ms is used in FLAIR.

For further details, see STIR.

**Related Articles:** Short tau inversion recovery (STIR), Inversion time (TI)

**Fluorescence**

*(Diagnostic Radiology)* Fluorescence is the emission of a photon from a material in response to excitation by an incident photon of higher energy. The wavelength of the emitted photon is characteristic of the fluorescent material.

A narrow light spectrum is emitted during fluorescence (with very short afterglow ~ ns). In medical physics, fluorescent materials are used mainly for PM detectors and image intensifier input screens (e.g. CsI(Tl)).

In contrast, broad light spectrum is emitted during phosphorescence (light continues after the radiation). In medical physics, phosphorescent materials are used mainly for monitor screens and image intensifier output screens (e.g. ZnCdS:Ag).

Materials exhibiting fluorescence in response to incident x-ray photons include zinc sulphide, calcium tungstate, barium lead sulphate and some rare earth compounds. The emitted wavelength from these materials is in the visible range.

Fluorescent materials are employed in the screens used in film/screen combinations to convert incoming x-ray photons to visible light photons that in turn blacken the film.

**Fluorescent screen**

*(Diagnostic Radiology)* In film screen x-ray imaging, the film is in contact with an intensifying fluorescent screen. The purpose of

**FIGURE F.35** The screen absorbs x-ray photons and emits photons in the visible range that blacken the film.

the screen is to convert incident x-rays to visible light that in turn blackens the film (Figure F.35). The screens are made from materials which exhibit fluorescence, typically rare earth phosphors. The phosphors have high atomic numbers and absorb x-rays much more efficiently compared with film alone. Only a very small fraction of the latent image on the film is formed directly by the x-ray beam – nearly 99% is due to light photons emitted from the intensifying screen.

Screens are built into the light-tight cassette used to hold the film. Usually, there is a screen on both sides of the film. On closing the cassette, the screens come into close contact with the film surface. Use of screens on either side of the film yields better resolution than a single screen of the same total thickness. Where the efficiency of conversion of x-ray energy to light (the screen’s conversion efficiency) is very high, the upper screen may be thinner than the lower screen to avoid overexposure by the upper screen relative to the lower screen.

The ‘speed’ of a film screen combination is a measure of its sensitivity to x-ray exposure. Increased screen thickness, increased phosphor size and increased phosphor efficiency increase the film screen speed. Resolution is predominately a function of the screen rather than the film, as the film has much higher resolution than the screen. A thicker screen, larger phosphor size and poor film to screen contact all reduce resolution.

Materials used as screen phosphors include rare earth phosphors (e.g. lanthanum oxybromide, gadolinium oxysulphide) and calcium tungstate. Rare earth screens provide higher speed than calcium tungstate. Calcium tungstate and lanthanum screens emit in the blue region of the spectrum, while gadolinium phosphors emit in the green.

Nearly all x-ray interactions with the phosphor material are photoelectric. The number of x-ray photons absorbed increases suddenly for x-ray energies above the K-edge of the phosphor. The K-edge of a rare earth screen (39 keV for lanthanum, 50 keV for gadolinium) is lower than that of a calcium tungstate screen (70 keV). For all x-ray energies between the K-edges of rare earth and calcium tungstate, the rare earth screen absorbs x-rays more efficiently. For the range of x-ray energies used in diagnostic radiology, rare earth screens are more efficient, absorbing 50%–60% of the incident beam, compared with 20%–40% for a calcium tungstate screen.

**Fluorescent x-rays**

*(General)* Fluorescent x-rays are x-rays emitted from a material in response to excitation of electrons in some manner. Fluorescent x-rays are characteristic of the material, and result from the same process of electronic transitions that produce characteristic x-rays in an x-ray tube. Usually, the term fluorescent implies that the excitation source is high-energy photons such as gamma rays or x-rays rather than a stream of electrons.
An incident high energy photon with sufficient energy can eject an inner-shell electron. The vacancy resulting from ejection of the electron is filled by an electron falling from a higher shell. The excess energy between the two levels is emitted as a photon.

The energy gap between any two electron levels is characteristic of the atom. In tungsten, for example the primary transition from the L to K shell results in emission of an x-ray with an energy of 59.3 keV (Curry et al., 1990). There is more than one energy associated with each electron shell, and transitions to a vacancy in an inner shell can take place from a number of shells. This results in a number of characteristic lines in the energy spectrum for a particular atom (Figure F.36).


**Fluorescent yield**

*(General)* In an electronic transition from a higher shell to a vacancy in a lower shell, energy is released. The energy may be released as a characteristic photon (an x-ray if energetic enough) or an Auger electron. In Auger emission, the energy difference between the two levels is imparted to an outer shell electron which is ejected from the atom (Figure F.37).

The fluorescent yield ($\omega$) is the probability than a transition results in emission of a photon. The probability that the transition results in emission of an Auger electron is $1 - \omega$.

**Fluorine**

*(General)*

<table>
<thead>
<tr>
<th>Symbol</th>
<th>F</th>
</tr>
</thead>
<tbody>
<tr>
<td>Element category</td>
<td>Halogen</td>
</tr>
<tr>
<td>Mass number $A$</td>
<td>19</td>
</tr>
<tr>
<td>Atomic number $Z$</td>
<td>9</td>
</tr>
<tr>
<td>Atomic weight</td>
<td>18.998 g mol$^{-1}$</td>
</tr>
<tr>
<td>Electronic configuration</td>
<td>1s$^2$ 2s$^2$ 2p$^5$</td>
</tr>
<tr>
<td>Melting point</td>
<td>53.53 K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>85.03 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>1.7 g L$^{-1}$ (gas)</td>
</tr>
</tbody>
</table>

**History:** Fluorine was first identified as part of the compound fluor spar (calcium fluoride). This was initially described in 1530 by Georgius Agricola in its capacity of catalyst in the compounding of various metals. It was also found to be useful in the process of etching glass, when combined with acid. Due to its extremely high reactivity, the highest of any known element, it proved difficult to isolate in its elemental form, but this was finally achieved in 1886 by Henri Moissan, by electrolysis of hydrofluoric acid.

**Isotopes of Fluorine:** A 100% of naturally occurring fluorine is found as stable $^{19}$F. It is an extremely reactive gas and is seldom found in its elemental form, which exists as F$_2$, but readily forms compounds with many other elements. Radioactive $^{18}$F is the most common of the PET imaging tracers and is used for a variety of different studies.

**Medical Applications:** PET imaging radioisotope – $^{18}$F decays by positron emission (97%) and electron capture (3%) to form $^{18}$O. Its high positron yield and a half-life of 110 min make $^{18}$F highly suitable for physiological imaging studies using PET. The fluorine radionuclide is usually introduced into the body as part of a larger molecule, such as fluorodeoxyglucose (FDG), and $^{18}$F-FDG is the most widely used radiopharmaceutical in clinical PET studies.

**Related Articles:** Positron emission tomography (PET), Fluorodeoxyglucose (FDG), Fluorine-18

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**FIGURE F.36** Ejection of an inner shell electron by an incident photon. Transition of a higher shell electron to fill the resulting vacancy results in emission of a photon with energy equal to the energy gap between the two shells.

**FIGURE F.37** (a) Electronic transition resulting in emission of a characteristic x-ray. (b) Electronic transition resulting in emission of an Auger electron.
Fluorine-18 (18F)
(Nuclear Medicine) Element: fluorine
Isotopes: 15 < N < 24
Atomic number (Z): 9
Neutron number (N): 9
Symbol: 18F
Production: Cyclotron, e.g. 18O(p,n)18F $\beta^-$

\[ 18F \rightarrow 18O \]

Daughter: 18O
Half-life: 109.77 min
Decay mode: $\beta^+$ (96.9%); EC (3.1%)
Radiation: $\beta^+$ (max 640 keV), annihilation photons
Photon energy: 511 keV (194%)
Dose rate from 1 MBq: 120 $\mu$Sv h$^{-1}$ at 30 cm (point source); 0.158 $\mu$Sv h$^{-1}$ at 1 cm (10 mL vial)
Absorption (range of $\beta^+$): 0.6 mm in tissue
Critical organ: lungs (inhalation), stomach wall (ingestion)
Critical organ dose: 0.03 – 0.057 mSv MBq$^{-1}$ (oral); 0.049 mSv MBq$^{-1}$ (inhalation)

Clinical Applications: Fluorine-18 is often produced by irradiation of 18O water using 10–18 MeV protons in a cyclotron. It is recovered as 18F sodium fluoride by passing the irradiated water target through a carbonate type anion exchange resin column, and the 18F is retained on the column matrix. It is eluted with potassium carbonate solution. 18F sodium fluoride is used for the synthesis of 18F fluorodeoxyglucose (FDG) as well as other 18F-labelled PET radiopharmaceuticals. It is also used for bone scintigraphy, since it localises in bone by exchanging with PO-4 ion in the hydroxyapatite crystal, resulting in PET images superior to 99Tc MDP scintigraphs.


Fluorodeoxyglucose
(Nuclear Medicine) 2-deoxy-2-[F-18]fluorodeoxyglucose (FDG) is a radiopharmaceutical used for PET imaging. FDG is a derivative of glucose, the predominant energy source for most cells of the body and for tumours. After intracellular phosphorylation via hexokinase, FDG-6-phosphate is not significantly metabolized and remains trapped in the cell.

Fluoroptic® probe
(General) ‘Fluoroptic’ is a registered trademark of LumaSense Technologies who make a range of fibre-optics-based temperature sensing transducers and meters.

Fibre-optic-based thermometers can have the advantage that the temperature sensor element may be made electrically non-conductive and be unaffected by electromagnetic interference or high magnetic fields. They may, therefore, be suitable for making measurements in harsh environments such as in MRI scanners.

Related Articles: Temperature control, Temperature probe

Fluoroscopic dose rate
(Diagnostic Radiology) Fluoroscopic systems can offer different level of operation (modes) – continuous fluoroscopy, pulsed fluoroscopy at a range of frame rates, different x-ray beam filtrations and FOV. The fluoroscopic dose rate is a parameter adopted to control the dose delivered to the patient, according to the selected mode.

Depending on the technical publication adopted (IEC, FDA, etc.), the dose rate reference points, the limits and the measuring conditions are different.

Fluoroscopic dose rate can be measured:

- At the entrance of the image receptor (without an anti-scatter grid), using a copper filter to simulate the patient (tolerance level: 0.8 $\mu$Gy s$^{-1}$ for an image receptor with a diameter of 25 cm; for alternate diameters, the limit is adapted according to the inverse square law)
- At the entrance of appropriate phantoms (usually Perspex to simulate the patient), in order to predict entrance skin dose to the patient (with a 25 cm thick phantom, the maximum skin dose, including backscattering, shall not exceed 100 $\mu$Gy min$^{-1}$)
- In air, 30 cm from the image receptor, using a lead filter to reach the maximum dose rate (the limit is 10 $R$ min$^{-1}$)
- 15 cm back from isocentre

Fluoroscopic portal imaging
(Radiotherapy) Portal imaging is the acquisition of images using a therapeutic x-ray beam in order to verify the beam delivery to the desired point of application during a radiotherapy treatment.

A common device commercially available is the fluoroscopic portal imaging: a metal plate is combined with a fluorescent screen to convert x-ray photons to light photons, which are viewed and digitised by a video camera.

The metal plate is typically 1–2 mm thick copper or steel or brass; for the fluorescent screen, a rare earth phosphor (usually gadolinium oxy sulphide) is generally adopted. A series of mirrors reflects the light photons into a lens and camera, which are surrounded by lead shielding.

The device enables imaging at any gantry angle and can be attached/detached or retracted from the position of use.

The images are digital and can undergo image processing, image matching and digital archiving. Moreover, the images are immediately available and can be used interactively to optimise field position during the treatment.

A recent developing application of electronic portal imaging device (EPID) is their use for pretreatment dosimetric verification.

Fluoroscopic timer
(Diagnostic Radiology) Fluoroscopic equipment is often used to perform very complex and time-consuming diagnostic and interventional procedures, which require special radiation protection measures. Fluoroscopic x-ray systems are always provided with
Fluoroscopy

(Diagnostic Radiology) Fluoroscopy is a mode of use of x-rays to visualise moving organs or to perform dynamic examination. This procedure requires a special x-ray detector. The earliest fluoroscopic equipment used just a fluorescent screen (phosphor) as a detector. This gives off a faint light only visible in a dark room. Later designs processed the image from the phosphor by a system of mirrors and lenses leading to 4 times minification and 16 times intensification of the image brightness. This image was then passed to a superorticon TV camera and viewed on a TV monitor. Contemporary fluoroscopic systems use an image intensifier (II), which allows a total amplification (gain) of the imaging system of up to 7000. The II is again linked to a TV camera and monitor. It is expected that in future the flat panel detectors with thin film transistors technology (TFT) will replace the II.

The image contrast in fluoroscopy depends on the x-ray spectrum exactly in the same way as in radiography. Due to this reason, various organs are filmed (static image) or observed in motion (fluoroscopy) using similar kV ranges. A major advantage of II is the greatly reduced intensity of the x-ray beam (normally an II system uses 0.3–3 mA anode current). However, the overall patient dose delivered during fluoroscopy is much greater than in radiography. This is due to the long examination times (several minutes).

Observation of hollow organs in fluoroscopy requires the use of various contrast media such as barium meal for imaging of the gastrointestinal system, or iodine contrast for imaging the blood vessels and the heart (angiography).

Related Articles: Image intensifier, Cinefluoroscopy

Fluoroscopy, digital

(Diagnostic Radiology) See Digital fluoroscopy

Fluoroscopy, mobile

(Diagnostic Radiology) See Fluoroscopy

Flux

(General) Flux is a measure of the rate of flow of some quantity through a defined surface area (Figure F.38).

Where the rate of flow of the quantity in question is defined by some vector field (V), flux can be defined as

\[ \text{Flux} = \int_{S} V \cdot dS \]

The formula represents integration of the component of the vector field aligned with the normal to each infinitesimal area (dS) over the entire area of the surface (S).

Flux density

(General) Flux density is a measure of the rate of flow of some quantity through a unit area of a defined surface (Figure F.38).

Where the rate of flow of the quantity in question is defined by some vector field (V), flux density can be defined as

\[ \text{Flux density} = \frac{\int_{S} V \cdot dS}{\int_{S} dS} \]

The formula represents flux divided by the total area of S.

Flying focal spot

(Diagnostic Radiology) In CT, a flying focal spot is used to increase sampling frequency and improve image spatial resolution. It is sometimes also referred to as a dynamic focal spot (Figure F.39).

Conventionally, on a third-generation scanner, the number of samples per projection equals the number of detector elements in the x–y plane. Some CT scanners employ a flying focal spot to double the number of samples per projection by sampling each detector element twice in every projection. To achieve this, the position of the focal spot on the anode is electromagnetically deflected (by deflecting the beam of thermal electrons from the cathode, so these hit a different spot on the anode). This deflection of the focal spot is made after the gantry has rotated a distance equal to half a detector element; after this, the focal spot returns to the position it was at for the previous projection measurement. This results in two interlaced measurements per projection.

The flying focal spot may also be available in the scan axis (z-axis) direction, such that two overlapping slices are acquired in each tube rotation, effectively doubling the sampling rate to half a detector width in that dimension.
fMRI (functional magnetic resonance imaging) (Magnetic Resonance) See Functional magnetic resonance imaging (fMRI)

Focal film distance (FFD)
(Diagnostic Radiology) FFD is the distance between the focal spot of the x-ray tube, or the radiation source, and the film, or the image receptor. It is also known as TFD or source image receptor distance (SID).

FFD is an important parameter influencing the intensity of the x-ray beam – the smaller the FFD, the greater the intensity of the x-ray beam.

This dependence is defined by the inverse square law – intensity of radiation decreases or increases by the square of the distance from the source.

Typical FFD values 80–110 cm

Doubling the distance from the x-ray tube to the patient (D) will cover four times the area at 2D but reduces the x-ray intensity to 1/4th \((1/2)^2\)

See the diagram in the article Target film distance (TFD).

Focal plane tomography
(Diagnostic Radiology) Focal plane tomography is another name for classical linear tomography. During the exposure, the tube and the image receptor are in synchronous movement, while the patient remains still (Figure F.40).

The resulting image shows the anatomical structures lying on the focal plane satisfactorily recorded, while the details above and below the focal plane appear blurred or eliminated.

The reason is that during the exposure in the focal plane the radiographic motion is minimum, and the region appears clearly defined on the image.

The level of the fulcrum can be selected according to the height of the anatomy of interest.

Furthermore, the section thickness is inversely proportional to the tomographic angle. The thinner the slice, the more detail for small region will be detectable.

The focal plane tomography was introduced in 1921 and was used intensively until 1980s.

More common diagnostic procedures were intravenous pyelograms, IVP (used to visualise abnormalities of the urinary system, including the kidneys, ureters, and bladder), temporal bone imaging and chest radiographic imaging.

Now, focal plane tomography is almost completely replaced by the computed tomography.

Related Article: Classical linear tomography

Focal plane tomography
(Nuclear Medicine) Focal plane tomography was used before emission computed tomography became widely available. Images were obtained at different projection angles and then shifted and superimposed so that one ‘longitudinal’ plane (parallel to the camera face) was focused whereas all other planes were blurred.

One example of focal plane tomography used a collimator with multiple pinholes. The different pinhole images were shifted and superimposed to bring certain planes into focus. Another example used a rotating slant-hole collimator.

The use of these techniques has mostly died out because underlying and overlying blurred planes obscure the focused plane of interest.

Related Article: Multiple pinhole

Focal point
(Ultrasound) For imaging purposes, a narrow ultrasound beam is desirable for resolution. The directivity of an ultrasonic beam can be altered by focusing. Although the general principles are the same as applied in optics, one must be aware of issues related to the wavelength and the situations when the dimensions of the transducer are comparable to the ultrasonic wavelength, as scattering will occur.

In this case, the laws of reflection and refraction still apply, but one might use the principle of superposition and finite elements to model the resulting US field. Focusing can be performed with curved transducers, focusing lenses or mirrors for a single element source or, as done in modern systems, with transducer arrays and electronic control of firing, delaying the firing of certain array elements in relation to others to emulate the effect of a curved transducer. The obvious advantage of the electronically controlled array is that the focal distance can be varied. The distance from the source to the focal point (or focus) is the focal length (F).

Wells (1977) points out that in the case of a spherical concave radiator, vibrating with uniform normal velocity and with a radius of the circular boundary, which is large in relation to both the wavelength \(\lambda\) and the depth of the transducer (Figure F.41), the ratio of the intensity of ultrasound at the centre of curvature to the intensity at the radiating surface is given by

\[
\frac{I_f}{I_0} = (kh)^2
\]

where \(k\) (the wave number) = \(2\pi/\lambda\) and that the point of greatest intensity is on the central axis but is not at the centre of curvature \(R\),
as one would expect. The focusing can be improved by suitable choice of dimensions (Wells 1997).


Focal spot
(Diagnostic Radiology) A focal spot is the area on the surface of an x-ray tube anode (target) that is bombarded by the high-energy electrons from the cathode and where the x-radiation is produced. Two focal spots can be defined – actual focal spot and effective focal spot (see the eponymous articles).

The actual focal spot is the area of the target bombarded by the accelerated thermal electrons. This area takes all thermal energy of the electron beam and this way determines the power of the x-ray tube. In classical x-ray tubes, this area is a projection of the shape of the filament coil (i.e. approximately rectangular shape). In some contemporary x-ray tubes (with electrical system which focuses the thermal electrons, e.g. Straton tube), the actual focal spot can vary in shape and size.

The effective focal spot is the projection of the actual focal spot down the central x-ray beam (towards the patient). Its size depends on the anode angle (see Line focus principle) and determines the resolution of the x-ray image. From optical point of view, the effective focal spot can be seen as the ‘x-ray point source’.

The relationship between the actual focal spot on the surface of the anode and the projected effective focal spot size is determined by the anode angle. The angle is a design characteristic generally in the range of 6°–24°. Small angles give the best relationship, large heat capacity and small projected size but limit the area that is covered by the x-ray beam (FOV) because of the heel effect. Figure F.42 shows a small section of a rotating anode with the different perspectives of a focal spot.

Related Articles: Anode, Cathode, Target, Line focus principle, Stationary anode, Focal spot actual, Focal spot effective

Focal spot, actual
(Diagnostic Radiology) The actual focal spot of an x-ray tube is the area of the target bombarded by the beam of thermal electrons. This area takes all thermal energy of the electron beam and this way determines the power of the x-ray tube (the larger the actual focus, the more powerful is the x-ray tube). In classical x-ray tubes, this area is a projection of the shape of the filament wire (coil). This way, it has approximately rectangular shape, where the width is roughly equal to the diameter of the filament coil (approximately 0.4 mm) and the height is determined by the length of the filament coil and the cosine of the anode angle. In some contemporary x-ray tubes (with an electrical system which focuses the thermal electrons), the actual focal spot can vary in shape and size.

Some sources distinguish two parts of the actual focal spot – electronic focal spot (the section of the electron beam bombarding the target) and thermal focal spot (the area of the target which has been bombarded – i.e. the actual focal spot).

In x-ray tubes with a rotating anode, the heat is distributed over a ring of the anode target, called thermal track (Figure F.43). In this case, the actual focal spot is often called ‘real’ (or ‘momentary’) focal spot (see article on Rotating anode). The thermal track distributes the temperature of the actual focal spot over a large area, thus effectively enlarging it (without enlarging the effective focal spot) and allowing an increase of power of the x-ray exposure.

Related Articles: Anode, Anode angle, Cathode, Target, Stationary anode, Focal spot effective

Focal spot, apparent
(Diagnostic Radiology) See Focal spot, effective

Focal spot, effective
(Diagnostic Radiology) The effective focal spot is the projection of the actual focal spot down the central x-ray beam (towards the patient). The effective focal spot determines the resolution of the x-ray image. From optical point of view, it can be seen as the ‘x-ray source size’. Some sources name the effective focal spot as apparent focal spot or as optical focal spot.

Ordinary x-ray tubes (for general radiography or fluoroscopy) have almost square effective focal spot with size on the order of 0.6–1.2 mm (a larger focal spot will produce too blurred images). Mammographic x-ray tubes (producing high resolution images) have similar effective focal spot with size on the order of 0.2–0.4 mm. Some contemporary x-ray tubes use electrical system which focuses the thermal electrons, thus producing variable effective focal spot shape and size.

The size of the effective focal spot \( F_e \) depends on the actual focal spot size \( F_a \) and the anode angle \( \alpha \) (see Line focus principle):

\[
F_e = \sin \alpha \cdot F_a
\]

Obviously, a small anode angle will produce small size of the effective focal spot. However, this will reduce the size of the useful x-ray beam (the section of the beam).

Because the effective focal spot is a projection down the central x-ray beam, its size is measured at the centre of the x-ray beam. Various tools are used for this purpose. The pinhole camera is used for direct measurement – Figure F.44. Other special test objects (e.g. star test or bar pattern) use high resolution patterns for indirect measurement – Figures F.46. All focal spot measurements require exact distances focus/test-object/film (specific for the test object or pinhole camera). Also, specific formulas are used for the calculation of the effective focal spot size from the image of the test objects with high resolution patterns.

Figure F.45 shows a bar pattern test object used for indirect measurements of the effective focal spot size. In this case, the size of the focal spot is derived (through a table) from the size of the blurred patterns. Note that here two different patterns are blurred (the 8th and 9th patterns) showing that the focal spot is not square.

In a large x-ray field, the effective focal spot size will increase its size towards the periphery of the field (away from the centre).
This phenomenon creates varying spatial resolution of the images (the best one being at the centre of the x-ray beam). This variation is most prominent in the cathode-anode axis of the tube, where the Heel effect has an additional influence to the effective focal spot size and x-ray beam intensity – Figure F.45. Note – the figure presents only a rough idea about the effective focal spot size and shape at the edges of the maximal x-ray field (at least 1 m away from the x-ray tube). Aiming at better understanding, the central spot of the effective focal spot size on Figure F.46 has been shown as rectangular with significantly different sides.

Figures F.47 and F.48 show real images of measurement of the effective focal spot made with multiple pinhole camera. On both images, the vertical direction is the cathode-anode direction. In Figure F.47, the measurements are from a new modern x-ray tube (production 2007); in Figure F.48, the measurements are from an old x-ray tube.
Focal spot selection

(Production 1986). Note that the new x-ray tube presents overlapping images of both the fine focus (the smaller darker spot) and the broad focus (the larger paler spot). Both foci of the old x-ray tube do not overlap (the fine is below the broad focus). The image of the new x-ray tube shows better the enlargement of the effective focal spot.

Related Articles: Anode, Anode angle, Anode heel effect, Cathode, Line focus principle, Stationary anode, Focal spot actual, Pinhole camera

Focal spot selection

(Diagnostic Radiology) The focal spot selection is made through the filament circuit of the high voltage generator.

Each focal spot is associated with a different filament wire. The selection takes into consideration the maximum permissible temperature of the filament of the respective focal spot.

The current through the cathode filament (filament current \( I_f \)) depends also on the maximum permissible power for selected focal spot \( P_{\text{max}} \). The filament circuit assures anode current \( I_a < P_{\text{max}}/U_a \) for each selection of focal spot and anode voltage.

The selection of a focal spot size for a specific clinical procedure generally represents a compromise between two requirements. Small sizes are desirable to minimise image blurring and provide good visibility of anatomical detail and small signs of pathology. However, large focal spot areas are required to reduce heat concentration and permit the necessary x-ray exposure to be produced for the procedure.

Selection of focal spot size occurs in two stages. The first is the selection of the tube design to be used for the various clinical procedures. Tubes with small focal spots are selected for procedures that require high visibility of detail such as mammography. Tubes with larger focal spots are selected for procedures that require the production of relatively high exposures to penetrate and image large body sections such as the abdomen and the lateral chest.

Most tubes are designed with two focal spot sizes. The operator or the automatic exposure control function of the equipment select between the ‘small’ and the ‘large’ focal spot size for a specific procedure. Generally, the smaller size is used for good image detail if it has the necessary heat capacity.

Related Articles: Cathode, Filament, Filament circuit, High voltage generator, Target, Focal spot actual, Focal spot effective

Focal spot selector

(Diagnostic Radiology) See Focal spot selection

Focal zone

(Ultrasound) In transmit mode, a number of separate transmissions must be used if lateral resolution is to be maintained over the entire imaging depth. For each transmission, the focus is placed at different depths, and only the part of the received data that is close to respective focus (referred to as the focal zone) is used for image data. In Figure F.49, three transmissions are used to make up the image. Note how the aperture is increased with increasing imaging depth to maintain the lateral resolution (F-number).

In Figure F.49, only the part of the received data that is close to respective focus is used for image data. Note how a larger aperture is used for the deeper zones.

Focused collimator

(Nuclear Medicine) See Diverging collimator

Focused grid

(Diagnostic Radiology) A focused grid is designed so that the spaces between the attenuating strips are at an angle so that they align with or focus a point in space as shown in Figure F.50.
Focusing cup

(Diagnostic Radiology) This is a small metal cup which holds the cathode wire of the x-ray tube. Its function is to focus the thermal electrons emitted by the cathode filament, which otherwise produces quite spread beam of electrons towards the anode. This way, the focusing cup minimises the focal spot size over the anode. The focusing cup is specially shaped and is made of molybdenum, nickel or steel, because of their poor thermionic emission. Placing the filament wire inner or outer in the cup (during the production of the x-ray tube) changes the focusing of the electron beam. When negative voltage is applied to the focusing cup, it becomes a Wehnelt electrode (see the eponymous article). Changing the voltage of this electrode changes the shape of the beam of thermal electrons, hence the focal spot size. This electrode can also stop the beam of electrons (in the grid-controlled x-ray tubes).

Related Articles: Cathode, Wehnelt electrode, Focal spot, Grid-controlled tube

Foetus

(General) In humans, the foetus is the unborn young from the end of the 8th week after conception to the moment of birth, as distinguished from the earlier embryo. In a foetus, all major body organs are present.

See also Embryo

Fog

(Diagnostic Radiology) See Film fog

Foot switch

(Diagnostic Radiology) See Dead man’s switch

Forbidden energy gap

(General) All atoms normally have electrons surrounding them; however, there are only a fixed number of possible stable orbits or shells where they can exist. Electrons may move from inner orbits to outer orbits or be ejected completely if they are given enough energy through absorbing electromagnetic radiation. Low orbits require more energy to escape, whilst outer orbits need less.

The limited number of orbits means that the electrons around atoms have a clearly defined ability to absorb/re-emit radiation only at certain fixed energies equaling the energy difference between allowed orbits. Above this layer of valence electron orbits lie further unfilled orbits to which excited electrons may be energised to fill. This defines the typical bands seen in optical absorption/emission spectra of elements in gaseous form.
In solids, the nearness and binding forces between atoms and the possible mix of elements result in these individual energy levels being reduced to allowable energy bands:

The forbidden energy bands of some solids means that in some cases the valence bands (normally filled with electrons) and the conduction bands (normally empty of electrons) are separated, trapping the electrons to their individual atoms and preventing them from easily moving from atom to atom.

In semiconductors, the energy needed to cross the forbidden gap is only low, and this can be overcome with low electric potentials, leading to the use of semiconductors in electronic devices.

Insulators have wide forbidden bands, which can only be overcome with strong electric fields, leading to common electrical insulators.

Common electrical conductors such as copper, silver and gold have no forbidden band, and their electrons can freely move in and out of the valence band, providing easy electrical current flow with very little electric field needed.

It is interesting to note that while conductors have a resistance to current flow which rises slightly with temperature, one of the defining properties of semiconductors is that their electrical resistance decreases significantly with increasing temperature.

**Force balance**

(*Ultrasound*) Force balances are used to measure the power output of ultrasound devices. The total acoustic power passing through an area is defined by the integration of the intensity over that area. This can be expressed as a summation:

\[ W = \sum I_{\text{w}} \cdot \Delta A \]

where

- \( I_{\text{w}} \) is the temporal average intensity
- \( \Delta A \) is the square of the sample distance

The summation is performed for all \( I_{\text{w}} \) (>1% of the peak) measured in the specified plane. This parameter can be measured with a hydrophone system but this is complex and time-consuming. The method usually preferred is to use a force balance.

The ultrasound beam exerts a force on the balance. There are two basic designs of force measurement device, using either absorption or reflection of the ultrasound beam (*Figures F.52 and F.53*).

For a plane wave incident on a perfectly absorbing target, the radiation force is given by

\[ F = \frac{IA}{c} = \frac{W}{c} \]

where \( c \) is the speed of ultrasound in the propagation medium, usually water.

It may be difficult to achieve perfect absorption, and an alternative method is to use a conical air-filled reflector. For a perfect reflector, the force and power are related by

\[ W = \frac{cF}{2\cos^2\theta} \]

where \( \theta \) is the angle between incident beam and the normal to the reflecting surface.

The forces that result from diagnostic and therapeutic devices are small, typically mN. Sensitive weighing devices are used to measure the slight change in weight. To improve sensitivity, the target may be suspended on a separate structure linked to the balance to remove the weight of the water and container (*Figure F.54*).

**Related Articles:** Acoustic power, Acoustic pressure, Absorption, Intensity
Force, electrostatic

(General) The magnitude of the electrostatic force \( F \) on a charge \( q_1 \) due to the presence of another charge \( q_2 \) is given by

\[
F = k_e \frac{q_1 q_2}{r^2}
\]

where
- \( r \) is the distance between the two electric charges \( q_1 \) and \( q_2 \)
- \( k_e \) is proportionality constant

The proportionality constant \( k_e \) is called Coulomb’s constant and is given by

\[
k_e = \frac{1}{4\pi\varepsilon_0} = \frac{\mu_0\varepsilon_0^2}{4\pi}
\]

\[
= 8.9875517873681764 \times 10^9 \text{ N m}^2 \text{ C}^{-2}
\]

where
- \( c_0 \) is the speed of light in a vacuum equal to 299,792,458 m s\(^{-1}\)
- \( \mu_0 \) is the magnetic constant defined as \( 4\pi \times 10^{-7} \text{ H m}^{-1} \)
- \( \varepsilon_0 \) is the electric constant equal to \( 1/(\mu_0 c_0^2) \sim 8.854187817 \times 10^{-12} \text{ F m}^{-1} \)

Related Articles: Charge, Coulomb


Forces, nuclear

(General) See Nuclear forces

Forward treatment planning

(Radiotherapy) Forward planning is a conventional radiotherapy treatment planning technique in which the planner selects by experience the required number of open and/or wedged treatment beams of appropriate beam geometries. He then calculates either manually or by means of a treatment planning computer the composite distribution of dose by adding the dose contributed by each of the treatment beams. If the dose and the distribution of dose in the irradiated volume are unsatisfactory, the planner varies the beam parameters and geometries and repeats the calculation. The processes are repeated until an acceptable treatment plan is achieved.

Four-rectifier circuit

(Diagnostic Radiology) Four-rectifier circuits, also known as bridge rectifier (or Graetz circuit), consist of four diodes connected in a way to form a bridge. This arrangement allows full wave rectification by converting the input AC voltage into a positive pulsating DC voltage (Figure F.55).

See Rectifier

Fourier reconstruction

(Generic) See Filtered back projection

Fourier spectrum framework

(Magnetic Resonance) In MRI, there is a direct relationship between the collected data and the reconstructed image through Fourier theory and the Fourier transform. The same relationship is used in nuclear magnetic resonance (NMR) spectroscopy which is taken as the example here since it is limited to one dimension.

Briefly, signal from a specified volume of tissue is acquired after the transmission of an RF pulse (excitation). This yields the FID, i.e. signal as a function of time. After applying the Fourier transform to the detected signal, a spectrum revealing its frequency content is obtained. Mathematically, this can be considered a decomposition of the initial signal (time domain) into an infinite sum of sines and cosines with different frequencies (frequency domain). The signal as a function of time and the frequency spectrum is a Fourier pair and the signal can be written as the inverse Fourier transform of the frequency spectrum:

\[
f(t) = \int F(u)e^{i2\pi ut} du
\]

where
- \( f(t) \) is the FID
- \( u \) is the frequency variable
- \( F(u) \) is the complex data represented in the frequency domain

This can also be written as

\[
F(u) = |F(u)|e^{i\theta(u)}
\]

The magnitude \( F(u) \) is called the Fourier spectrum and \( \theta(u) \) is the phase angle.

In spectroscopy, a spectrum allows visualisation of the relative occurrence of various metabolites, which, due to their varying chemical environment, contain protons with correspondingly varying resonance frequencies.

Related Articles: Fourier transform, Free induction decay, Frequency spectrum
**Fourier transform**

*(General)* Fourier transform refers to the process of decomposing a function in terms of the sum of sinusoidal functions with different frequencies. The Fourier transform can also refer to the function’s representation in the frequency domain.

To read more about Fourier analysis and Fourier series, follow the external links.

**Hyperlinks:** [http://mathworld.wolfram.com/FourierTransform.html](http://mathworld.wolfram.com/FourierTransform.html)

**FOV (field of view)**

*(Nuclear Medicine)* See Field of view (FOV)

**Fractional anisotropy (FA)**

*(Magnetic Resonance)* FA is a scalar index for diffusion anisotropy. It is determined from the eigenvalues of the diffusion tensor and is defined as

$$\text{FA} = \sqrt{\frac{3\left(\lambda_1 - \lambda\right)^2 + \left(\lambda_2 - \lambda\right)^2 + \left(\lambda_3 - \lambda\right)^2}{2\left(\lambda_1^2 + \lambda_2^2 + \lambda_3^2\right)}} = \sqrt{\frac{3\left(\sum \lambda_i^2 - \lambda^2\right)}{2\sum \lambda_i^2}}$$

where

- $\lambda_i$ is the $i$:th eigenvalue of the diffusion tensor
- $\lambda$ is the mean of the eigenvalues, i.e. the mean diffusivity (MD)

The FA ranges from 0 (isotropic diffusion) to 1 (anisotropic diffusion). The FA is a rotationally invariant metric, i.e. it is independent of the major diffusion direction and the orientation of the object (Figure F.56).

**Related Articles:** Diffusion tensor, Relative anisotropy

**Fractionation**

*(Radiotherapy)* Generally, radiotherapy treatment is delivered once a day, 5 days a week for up to 8 weeks. Each daily treatment is called a fraction and such regimes have been shown to be clinically effective and acceptable, giving a favourable therapeutic effect in most cases: better tumour control is obtained for a given level of normal tissue toxicity when the radiation dose is fractionated rather than delivered as a single dose. The effectiveness of fractionated radiotherapy can be understood by consideration of the 5Rs of radiobiology. Dividing the radiotherapy dose into fractions spares normal tissues since the cells can repair some of the radiation damage in the time between fractions, and cell repopulation will occur provided the overall time is sufficiently long. Conversely, fractionating the dose increases the damage to the tumour due to reoxygenation and the redistribution of cells into radiosensitive phases of the cell cycle between fractions.

Many of the regimens currently in use have developed as a result of expediency rather than from radiobiological principles. Historically, the most common dose per fraction in use has been 2 Gy. However, in the Northern United Kingdom, fraction sizes of 2.67–2.75 Gy per day have been used for many years, introduced largely to ease the burden on thinly spread resources. It should be noted that the total dose delivered with these hypofractionated regimens is reduced and that the clinical results have generally been shown to be comparable to those for the more standard 2 Gy per fraction protocol. Indeed, there is renewed interest in hypofractionation in the light of new data on the alpha-beta ratio values for some tumours, the results of fractionation trials such as START and the advent of intensity-modulated radiation therapy. More details can be found in the article on Hypofractionation.

As discussed earlier, the prolongation of treatment by the use of fractionation has been shown to be beneficial in many cases. However, the excessive prolongation of treatment may actually have a detrimental effect on the therapeutic efficacy of the treatment if proliferation of tumour cells becomes significant. Irradiation of tumour cells can trigger the surviving cells to divide faster than before. This is known as accelerated repopulation (further details can be found in the article on Repopulation). Hyperfractionation, particularly accelerated hyperfractionation, offers a possible solution to this problem. Continuous hyperfractionated accelerated radiation therapy (CHART), which reduces overall treatment from 6–7 weeks to 12 days and gives 36 small fractions, has been tested in multicentre randomised controlled clinical trials. The trial in non-small-cell lung cancer showed improvement in survival, and this regimen is now the government-recommended standard of care for eligible patients in the United Kingdom. More details can be found in the article on Hyperfractionation.

In the United Kingdom, the Royal College of Radiologists published a report in 2006 in which they identified fractionation regimens for which there is high-quality evidence for both safety and efficacy. They found a state of equipoise in many clinical situations, meaning that there is genuine uncertainty over whether or not the treatment will be beneficial, where published evidence was insufficient to favour one particular regimen over another and affirmed that clinical trials are required to resolve these issues.

In cases where a change in fractionation regimen is considered, perhaps due to an interruption of treatment or as part of a clinical trial, it is useful to be able to compare the regimens in terms of the effect on both the tumour and the normal tissues. This is possible by using the linear-quadratic model to evaluate the biological effective dose (BED). In the case of normal tissues, a more useful parameter may be the equivalent total dose in 2 Gy fractions (EQD2) since most clinical experience of normal tissue tolerance has been obtained from 2 Gy per fraction regimens. More detail can be found in the article on BED.

**Abbreviations:** BED = Biological effective dose and EQD2 = Equivalent total dose in 2 Gy fractions.

**Related Articles:** Alpha beta ratio, Biological effective dose, Cell cycle, Fractionation, Interruption of treatment, Linear quadratic (LQ) model, Radiosensitivity, Repair,
Fractions

(Radiotherapy) Generally, radiotherapy treatment is delivered once a day, 5 days a week for up to 8 weeks. Each daily treatment is called a fraction. For more information on the rationale of fractionation, see the articles on Fractionation and The 5Rs of radiobiology.

Related Articles: Fractionation, 5Rs of radiobiology

Frame mode for digital image acquisition
(Diagnostic Radiology) See Acquisition modes for digital image

Frame rate
(Nuclear Medicine) The frame rate is the frequency by which an imaging system produces new unique consecutive images. The frame rate is normally expressed in frames (or images) per second or in hertz (Hz).

Frame rate
(Ultrasound) The frame rate is the nominal frequency at which an ultrasound image is updated. It applies to B-mode and colour flow imaging and is usually indicated on the screen image in units of Hz or fps (frames per second). The time taken to image an area is dependent on the number of pulses used for each frame and the pulse repetition frequency. This in turn depends on a number of factors, for example whether B-mode or colour flow imaging is used, the depth and width of the image, line density and number of pulses used for each ‘line’ of ultrasound (e.g. if multiple transmit zones are used). It is possible that the colour flow and B-mode frame rate may differ in an image where both are displayed and updated. Frame rates typically range from 50 Hz for superficial B-mode images to 4 Hz for deep structures with a large area of colour flow imaging. Manufacturers go to considerable lengths to offset other parameters (e.g. line density) to ensure that frame rate remains acceptable for the application.

Fraunhofer zone
(Ultrasound) See Diffraction

Free-air ionisation chamber
(Radiotherapy) One of the most significant applications of ionisation chambers is in the measurement of the dosimetric quantity exposure in a photon beam. An air-filled ionisation chamber is particularly well-suited for this purpose because the exposure is defined in terms of the amount of ionisation charge created in a mass of air. Due to the exposure definition, its measurement at a particular point requires the collection of all the ions of any one sign generated in a known mass of free air contained in a volume placed around that point. In practice, it is needed to follow over their entire range each secondary electron created in the sensitive volume of the ionisation chamber by the photon interaction and to measure all the ionisations created along their track. As the range in air of the produced secondary electrons can be very large, it is impractical to design an instrument that permits a direct measurement and therefore a principle of compensation is used. If the sensitive volume of the ionisation chamber is surrounded by a sufficient thickness of equivalent air which is also subject to the same exposure, an exact compensation will occur as the ionisation charge created outside the volume, from secondary electrons which were formed within the volume, is precisely balanced by the charge created within the volume from secondary electrons produced in the surrounding air. This condition is called CPE. This situation is shown in Figure F.57.

The design of the ionisation chamber based on this compensation is schematically shown in Figure F.58 and the ionisation chamber is called free-air ionisation chamber.

The incident radiation is collimated so that it passes between the electrode system of a parallel plate ionisation chamber. The system consists of a plate maintained at a high potential and a collecting plate surrounded by earthed guard rings. The guard rings are necessary to ensure that the electric field lines are perpendicular to the electrodes in the sensitive volume space, avoiding lateral field distortion.

To make all the secondary electrons generated in the sensitive volume complete their tracks in air, the volume should be surrounded by an air thickness greater than the range of the most energetic secondary electrons. In practice, one prefers to measure the exposure \( E \) at the location where the entrance port is given by

\[
E = \left( \frac{d'}{d} \right)^2 \frac{q}{\rho A' L} = \frac{q}{\rho AL}
\]

where
- \( d \) is the distance of the entrance port from the source
- \( d' \) is the distance from any point in the source
- \( q \) is the collected charge
- \( \rho \) is the density of air
- \( L \) is the length of the collecting plates
- \( A' \) and \( A \), respectively, are the cross-sectional area at a distance \( d' \) and \( d \)
The distance from the entrance port to the sensitive volume of the ionisation chamber should be not so large to attenuate the photon beam significantly.

In Table F.2, the minimum air equivalent thickness required to establish the electronic equilibrium condition is reported. The thickness is based on the range of secondary electrons in water.

The free-air ionisation chamber is the primary standard for air kerma in air for superficial and orthovoltage x-rays produced by applied voltage up to 300 kV. The free-air ionisation chamber cannot function as a primary standard for photons beams of higher energy because the air thickness surrounding the sensitive volume needed to establish the electronic equilibrium condition in air would become very large. This would make the chamber very bulky and also the various required corrections and their uncertainties would become problematic.


Free induction decay (FID) (Magnetic Resonance) The application of a 90° excitation pulse rotates the net magnetism from the z to the y direction. The spins will then precess about the z direction in the transverse plan xy at an angular frequency $\gamma B$, where $B$ is the field that the individual nuclear spins experience and $\gamma$ is the gyromagnetic ratio. They will also simultaneously relax to alignment along the z direction. If there is a current loop in a plane parallel to the x, z plane, the changing magnetic flux in the loop will induce a current of frequency $\omega_0$ in this loop as shown in Figure F.59.

The resulting signal, an exponentially damped sine wave, is called FID because the spins begin to precess freely, the signal starts to decay with time and the spins induce a current in the receiving coil. In Figure F.60, the FID signal is shown.

The signal is of the form $\exp(-t/T_{2}^{*})\cos(\omega_0t)$, where $T_{2}^{*}$ is the total transverse relaxation time. Two effects contribute to the decay rate $1/T_{2}^{*}$. The first effect is the spin-lattice relaxation process representing the return of magnetisation to the z direction while the second effect is because the different nuclei in different environments and locations experience different fields differing by $\Delta B$ from the mean. $T_{2}^{*}$ is always less than $T_2$ and the relation between $T_2$ and $T_{2}^{*}$ is $1/T_{2}^{*} = 1/T_2 + \gamma \Delta B$. $T_{2}^{*}$ also depends on diffusion, i.e. how rapidly spins spread out and leave the lattice.

### TABLE F.2

<table>
<thead>
<tr>
<th>Photon Energy (MeV)</th>
<th>Thickness (g cm$^{-2}$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.02</td>
<td>0.0008</td>
</tr>
<tr>
<td>0.05</td>
<td>0.0042</td>
</tr>
<tr>
<td>0.1</td>
<td>0.014</td>
</tr>
<tr>
<td>0.2</td>
<td>0.044</td>
</tr>
<tr>
<td>0.5</td>
<td>0.17</td>
</tr>
<tr>
<td>1</td>
<td>0.43</td>
</tr>
<tr>
<td>2</td>
<td>0.96</td>
</tr>
<tr>
<td>5</td>
<td>2.5</td>
</tr>
<tr>
<td>10</td>
<td>4.9</td>
</tr>
</tbody>
</table>

FIGURE F.58 Schematic diagram of the free-air ionisation chamber.

FIGURE F.59 A magnetic moment rotating in the $xy$ plane induces a voltage in a coil in the $yz$ plane.
Free radicals

(General) A free radical is an atom or molecule with at least one unpaired orbital electron in its outer shell. This results in a highly reactive state.

In the body, free radicals usually occur on oxygen molecules and the two most important are the superoxide and hydroxyl radicals. Superoxide has the chemical formula $O_2^-$ and is biologically toxic. Indeed, it is used by the immune system to kill invading micro-organisms. The hydroxyl radical (OH) is the neutral form of the hydroxide ion and is highly reactive with a half-life in the body of around $10^{-9}$ s. Both types of free radical may participate in unwanted reactions resulting in cell damage, but it is estimated that around two-thirds of x-ray damage to DNA in mammalian cells is caused by the hydroxyl radical. Reactions between free radicals and DNA resulting in mutations have been linked to many forms of cancer. Free radicals have also been linked to many diseases, including Parkinson’s disease and Alzheimer’s, and to the aging process.

Abbreviation: DNA = Deoxyribonucleic acid.


Frequency

(Ultrasound) The frequency ($f$) of an ultrasound wave is the number of compressions (or rarefactions) per second, Figure F.61. The unit for frequency (1/s) is Hertz (Hz). Ultrasound is a sound with a frequency higher than 20 kHz. In diagnostic imaging, frequencies in the range of 2–20 MHz are most often used. High-frequency ultrasound contributes to high-resolution images but reduces penetration depth as attenuation increases with higher frequencies. In modern scanners, the frequency of the transmitted pulses is adjusted automatically depending on the transducer and the type of diagnostic option that have been chosen.

A period ($T$) is the time it takes for a signal to repeat itself and $T = 1/f$, Figure F.62. The acoustic wavelength ($\lambda$) is the distance between two corresponding points on the wave (e.g. two compression peaks), and as the wave travels with speed of sound ($c$), the relationship between $\lambda$ and $f$ can be written as $\lambda = cf$.

Related Article: Wavelength

Frequency encoding

(Magnetic Resonance) Frequency encoding is one of the principal methods used for spatial localisation in MRI. It is normally used in conjunction with phase encoding to localise signal within a selected slice.

Frequency encoding is achieved by applying a static field gradient while the NMR signal from the slice is being collected (normally in the form of an echo). In the presence of this gradient, elements of transverse magnetisation at different locations along the direction of field variation precess at different frequencies, according to the equation $\omega(x) = \gamma B = \gamma(B_0 + G_x x)$, where $x$ is displacement along the gradient direction, $G_x$ is the gradient amplitude (in mT m⁻¹), $\gamma$ is the gyromagnetic ratio of the nucleus (normally $^1$H) and $B_0$ is the strength of the static magnetic field. The resulting signal can
Frequency spectrum

A representation of a variable, for example signal intensity as a function of frequency is denoted a frequency spectrum. See also Fourier spectrum framework.

Related Articles: Fourier spectrum framework, Fourier transform, Free induction decay

Frequency-tailored RF pulse

(Magnetic Resonance) A frequency-tailored RF pulse contains energy only within a specified frequency range. This is important for slice excitation or for frequency-selective suppression pulses. The frequency range of the RF pulse can be tailored (designed) by manipulating the shape and the duration of the RF pulse. The effect of an RF pulse on the spin population can be determined by the Bloch equations, and for the small flip angles, the Bloch equations can be approximated by the Fourier transformation of the RF shape in the time domain. The possibility to design the frequency profile of RF pulses is frequently used in MRI to obtain a manifold of important effects such the slice selection profile and selective excitation of fat tissue.

In the case of slice selection (see also Pulse sequence), the RF pulse is combined with the slice-selective gradient pulse. The slice thickness can be adjusted by changing the RF bandwidth (the selected frequency interval) and/or the amplitude of the slice-selective gradient. Slice-selective RF pulses have duration of a few milliseconds (ms), and their shape is designed to create, ideally, a rectangular slice profile and homogeneous flip angles over the whole width of the slice. Usually RF pulses with a sinc–Gaussian shape are used for this purpose.

Related Articles: Flip angle, Pulse sequence, RF pulse

Fresnel zone

(Ultrasound) See Diffraction

Fricke-based gel

(Magnetic Resonance) The use of iron (II) sulphate solutions for ionising radiation dosimetry has a lengthy history, having been originally proposed by Fricke in 1927. Irradiation of water generates free radicals, which convert Fe²⁺ ions into Fe³⁺ through a variety of reactions. The quantity of Fe³⁺ ions produced depends linearly on the radiation dose received. In 1964, Gore proposed incorporating Fricke solution into a gel matrix, so that incident radiation produces a stable spatial distribution of Fe³⁺ ions. This distribution can be interrogated using MRI and the dose distribution determined through the differential effect of Fe³⁺ and Fe²⁺ ions on water proton relaxation times.

The gel dosimetry system is usually based on gelatine or agarose (although other gels have also been used), which have the distinct advantage of tissue equivalent response to ionising radiation. The Fe²⁺ ions are normally supplied in the form of ferrous ammonium sulphate. Oxygenation is required to facilitate the free radical reactions, and there is an extensive literature regarding other possible additives to increase sensitivity.

The change in relaxation times is due to the greater paramagnetic moment of the Fe³⁺ ions as compared to Fe²⁺. Assuming that dipole interactions with the iron ions dominates water relaxation and that there is fast exchange of water molecules between the vicinity of the iron ions and bulk water, the change in the spin-lattice relaxation rate (Δ\(R\)) is directly proportional to the increase in Fe³⁺ concentration, and hence to dose. These changes in relaxation rate can be interrogated using any suitable \(T_1\)-mapping technique.

The advantage of gel dosimetry over other dosimetric techniques is that it can visualise complex radiation fields with high spatial resolution over a large volume. Anatomically realistic phantoms can be filled with gel, and inserts can readily be added to mimic bone, lung etc. The Frick-based gel is itself the dosimeter, so the method yields a continuous dose distribution in three dimensions without the need to introduce extraneous instruments. These advantages are particularly important for emerging conformal radiotherapy techniques where there may be abrupt variations in dose, and also in brachytherapy.
However, a number of complications have delayed the introduction of Frick-based gels for routine dosimetric use. Careful attention to manufacture and calibration is needed to achieve reproducible results. A dose of tens of Gy is generally needed to achieve significant changes in relaxation rate, with a fairly small region of linearity with dose before saturation at typically 50–75 Gy. Following irradiation, the imprinted dose distribution gradually degrades due to diffusion of iron ions, so there is a window of a few hours at most between irradiation and readout which requires unusually ready access to MRI equipment.

Some of these difficulties can be addressed through use of polymer-based gels, in which radiation-induced polymerisation brings about changes in relaxation rates. These gels are free of the diffusion problem associated with Frick-based gels, but problems with reproducibility mean that they too are not yet in routine clinical use.

**Related Article:** Relaxation


**Fricke dosimeter** *(Radiotherapy)* The standard Fricke dosimeter solution consists in 0.001 M FeSO₄ or Fe(NH₄)₂(SO₄)₂ and 0.8 M H₂SO₄ dissolved in air-saturated (~0.250 mM O₂) triple-distilled water. As the concentrations are small, the dosimeter can be considered to be effectively water. The conversion of ferrous to ferric ions induced by the radiation permits the use of the Fricke dosimeter as a secondary standard. The yield of the ferric ions in the solution is directly proportional to the absorbed dose and independent of the dose rate up to 10⁶ Gy s⁻¹. The principal reactions that yield ferric ion are

\[
\text{OH}^- + \text{Fe}^{2+} \rightarrow \text{OH}^+ + \text{Fe}^{3+} \\
\text{OH}^- + \text{Fe}^{2+} \rightarrow \text{H}_2\text{O}_2 + \text{Fe}^{3+} \rightarrow \text{OH}^- + \text{H}_2\text{O} + \text{Fe}^{3+} \\
\text{H}^+ + \text{O}_2 \rightarrow \text{HO}_2^- + \text{Fe}^{2+} \rightarrow \text{Fe}^{3+} + \text{HO}_2^-
\]

The ferric ion concentration can be measured by the change in the absorbance using a spectrometer at a wavelength of 304 nm at which the ferric ions absorb strongly and the ferrous ions do not absorb at all.

The absorbed dose to the dosimeter solution is calculated from the equation

\[
D = \frac{\Delta OD}{\varepsilon \cdot l \cdot \rho \cdot G(\text{Fe}^{3+})}
\]

where \(\Delta OD\) is the difference in optical density between the irradiated sample and the blank, \(\varepsilon = 2187\) L mol⁻¹ cm at 304 nm and 25°C, \(l\) is the actual path length in centimetres through the spectrometer cell, \(\rho = 1.024\) kg L⁻¹ for standard Fricke solution at 25°C and \(G(\text{Fe}^{3+}) = 1.607 \times 10^{-6}\) mol J⁻¹ is the radiation chemical yield of ferric ions at the irradiation temperature (\(t\)) for Co⁶⁰ γ rays and varies slightly with radiation energy. A correction for the temperature (\(t'\)) differing from 25°C was made according to the equation

\[
\varepsilon G_t = \varepsilon_{25} G_{25}(1 + k_1(25 - t))(1 + k_2(25 - t'))
\]

where

- \(t'\) is the irradiation temperature
- \(t\) is the spectrometer measurement reading temperature

The temperature coefficient \(k_1\) is approximately 0.0007°C⁻¹ and \(k_2\) is 0.0015°C⁻¹.

The useful range of the Fricke dosimeter is from about 4 × 10³ to 4 × 10⁶ cGy.

**Fringe field** *(Magnetic Resonance)* The magnet is one of the main components of the MR scanner. The magnet is typically a large cylindrical device that accommodates the patient during the MR. Its role is to provide a highly uniform and stable polarising field in which to carry out the scan. There are several different types of magnets: permanent, resistive and superconducting magnet. The magnets used in MRI not only generate a field in the required region but also produce a fringe field outside the magnet that affect a considerable volume. The extent of the fringe field produced by a magnet depends on the strength of the magnet and on its design. Permanent magnets are built of blocks of magnetic material and can generate fields up to about 0.3 T. Their fringe field is often of limited extent and rather small. Resistive magnets have a larger fringe field and internal fields up to about 0.15 T. Superconducting magnets represent the majority of manufactured magnets and can generate very high fields up to 10 T and more. They are a form of electromagnets which operate near absolute zero temperature, i.e. 4.2 K. The fringe field of these types of magnet depends upon the current in the

---

**TABLE F.3 Characteristics of Different Types of Magnet**

<table>
<thead>
<tr>
<th>Characteristics</th>
<th>Permanent Magnet</th>
<th>Resistive Magnet</th>
<th>Superconducting Magnet</th>
</tr>
</thead>
<tbody>
<tr>
<td>Field strength (T)</td>
<td>0.1–0.3</td>
<td>0.15–0.4</td>
<td>0.4–4</td>
</tr>
<tr>
<td>Distance to 0.5 mT fringe field (m)</td>
<td>&lt;1</td>
<td>0.5–2</td>
<td>3–10</td>
</tr>
<tr>
<td>Advantages</td>
<td>Negligible fringe field</td>
<td>Easy coil maintenance</td>
<td>High field strength</td>
</tr>
<tr>
<td></td>
<td>Low operating cost</td>
<td>Low capital cost</td>
<td>High field homogeneity</td>
</tr>
<tr>
<td>Disadvantages</td>
<td>Limited field strength</td>
<td>Potential field instability</td>
<td>Intense fringe field</td>
</tr>
<tr>
<td></td>
<td>Fixed field strength</td>
<td>High power consumption</td>
<td>High cryogen cost</td>
</tr>
<tr>
<td></td>
<td>Very heavy</td>
<td>Water cooling necessary</td>
<td>High capital cost</td>
</tr>
</tbody>
</table>
windings, the dimensions and design of the coil system and the presence of any ferromagnetic material in the environment (Table F.3). At sufficient distance, the fringe field \( (B_s) \) diminishes according to the dipole approximation as the cube of the distance from the centre of the magnet \( (r) \) and is proportional to the field strength of the magnet \( B_0 \):

\[
B_s \propto \frac{B_0}{r^3}
\]

Large distances are necessary, however, before the fringe field is reduced below the earth’s magnetic field \( (\sim 0.05 \text{ mT}) \). The profiles of equal fringe field intensity when sectioned by a plane appear as a series of loops usually termed isogauss lines. For a typical horizontal bore MRI magnet, these are illustrated in Figures F.65 through F.67, respectively, for 0.5, 1.0 and 1.5 T superconducting magnets on a horizontal plane across the isocentre (Table F.4).

Fringe fields can be substantially decreased through the use of a magnetic shielding. Shielding of magnets has the advantage of reducing the controlled area required around the magnet room. Passive shielding is obtained applying highly permeable materials to walls or to the surface of the magnet maintaining the needed homogeneity inside the bore. Active shielding is done by building a second magnet on the outside of the main magnet with a field of the opposite sign, thereby reducing the fringe magnetic field (Table F.5).

Operation of numerous electronic devices in use in hospitals (e.g. x-ray tubes, CRTs, scintillation cameras, and image intensifiers) may be affected by the fringe fields on the order of 0.1–5 mT. Effects on cardiac pacemakers have been reported for fringe fields as low as 1.7 mT. The most common effect was triggering of the asynchronous mode; the effect is very model and orientation dependent, and in the models tested, normal operation resumed when the pacemaker was removed from the field. Some pacemakers also exhibited significant torque when exposed. For this reason, current static magnetic field guidelines restrict exposures for wearers of cardiac pacemakers to 0.5 mT. This restriction is also applied to other implanted electronic devices and to prosthetic devices. The effect of the stray field on ferromagnetic (‘magnetic’) materials is a torque and an attractive force and with electric currents is to produce a torque on the conductor carrying the current. A ferromagnetic object is attracted to the area where the magnetic field has the strongest gradient, i.e. to the region where the field is most inhomogeneous and not to the region where the absolute field value is strongest. A long slender ferromagnetic object will be strongly pulled into the top of the magnet and will fall freely through the homogeneous region of the magnet and stick near the bottom exit port of the magnet. The object will move freely back through the

**FIGURE F.65** Isohypses (contour lines) for a 0.5 T magnet.

**FIGURE F.66** Isohypses (contour lines) for a 1.0 T magnet.

**FIGURE F.67** Isohypses (contour lines) for a 1.5 T magnet.
homogeneous region of the magnet but will be very difficult to push out to the top. This is because, even though the field is stronger in the centre of the magnet, it has less magnetic gradient in the homogeneous centre of the magnet than near the ends of the magnet where the flux lines spread out on their individual return path arcs. Obviously, the object will be a projectile which falls into the magnet, and, therefore, the described experience should not tested to avoid severe damages to persons and equipment.

Front pointer

(Radiotherapy) A front pointer is a device installed in the treatment head of a radiotherapy treatment machine, such as linear accelerator for accurate positioning of the radiotherapy treatment beam. The pointer indicates the centre of the radiation beam emitted from a radiotherapy machine. It is used to direct the radiotherapy treatment beam onto the correct treatment location on the patient during treatment set-up. The pointer can be in the form of a mechanical pointer or an optical pointer.

FSE

(Magnetic Resonance) See Fast spin echo (FSE)

Full width at half maximum (FWHM)

(Nuclear Medicine) This is a measure used to define the performance of a scintillation camera, in particular the spatial resolution and energy resolution of an imaging system. Consider a point source placed in the FOV of a detector. The resulting image will have a wider spatial spread due to the properties of the camera system. A count profile over the imaged point source is displayed in Figure F.68. The FWHM is the full width at half maximum, or \( A_{\text{max}}/2 \), expressed in the number of pixels or the distance in millimetres.

According to NEMA, ‘The full width at half maximum is the measure of the spread of a point or line spread function measured between locations 50% down on each side from the peak amplitude’.


---

**TABLE F.4**

<table>
<thead>
<tr>
<th>Fringe Magnetic Field (mT)</th>
<th>0.5T Axial/Transverse (m)</th>
<th>1.0T Axial/Transverse (m)</th>
<th>1.5T Axial/Transverse (m)</th>
<th>4.0T Axial/Transverse (m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.0</td>
<td>7.0/5.0</td>
<td>8.5/6.7</td>
<td>10/7.0</td>
<td>13/11</td>
</tr>
<tr>
<td>0.5</td>
<td>8.5/6.7</td>
<td>11/8.5</td>
<td>13/9.7</td>
<td>16/13</td>
</tr>
<tr>
<td>0.3</td>
<td>11/8.3</td>
<td>13/10</td>
<td>15/12</td>
<td>20/16</td>
</tr>
<tr>
<td>0.1</td>
<td>15/12</td>
<td>19/15</td>
<td>21/17</td>
<td>25/20</td>
</tr>
</tbody>
</table>

**TABLE F.5**

<table>
<thead>
<tr>
<th>Passive Shielding</th>
<th>Active Shielding</th>
</tr>
</thead>
<tbody>
<tr>
<td>Iron in the wall</td>
<td>Resistive shim coils</td>
</tr>
<tr>
<td>Iron around the magnet</td>
<td>Superconducting shim coils</td>
</tr>
<tr>
<td>Coi design</td>
<td></td>
</tr>
</tbody>
</table>

**FIGURE F.68** The FWHM is the curve width at half of the maximum signal.

**Full width at one-tenth maximum (FWTM)**

(Nuclear Medicine) This is a measure used to define the scintillation camera performance (e.g. the spatial resolution, the amount of scattered events and energy resolution of an imaging system). Consider a point source placed in the FOV of a detector. The resulting image will have a wider spatial spread due to degenerative properties of the camera system. A count profile over the imaged point source is displayed in Figure F.69. The FWTM is the full width at one-tenth maximum, or \( A_{\text{max}}/10 \) showing the amount of scattered events in the curve. The broader the FWTM, the larger the scattered events.

According to NEMA, ‘FWTM is the measure of the spread of a point or line response function measured between locations 90% down on each side from the peak amplitude’.

**FIGURE F.69** The FWTM is the curve width at one tenth of the maximum signal.
as a paramagnetic substance, it has a shorter $T_1^*$ than oxyhaemoglobin, all fMRI images are $T_1^*$ weighted. As gradient echo EPI (GE EPI) has a strong $T_2^*$ weighting and rapid acquisition times, it is an ideal pulse sequence for fMRI although other techniques such as spiral imaging, conventional gradient echo imaging and FSE imaging have also been used. Static field strength is another important parameter: the BOLD contrast increases with increasing field strength. For this reason, higher field strengths are often used for fMRI (Figure F.70).


**Full-wave rectification**

(Diagnostic Radiology) See Rectifier

**Functional magnetic resonance imaging (fMRI)**

(Magnetic Resonance) fMRI is a technique which is used to study brain areas which become active when the subject undertakes a particular task. Whilst structural MRI is used to study brain anatomy, fMRI is used to study brain function.

**Historical Background:** The development of fMRI during the 1990s was pioneered by Seiji Ogawa and Ken Kwong. Since then, fMRI has found clinical application in presurgical planning (to localise brain function). It has also been widely adopted as a popular and insightful tool in the fields of psychology, psychiatry, neurology and neuroscience. fMRI's rapid growth over the past decade can be attributed to two main factors: it is a non-invasive technique which is free from the risks associated with ionising radiation and it has good spatial resolution (up to the order of a millimetre).

**Mechanism:** fMRI is usually based on blood oxygen level–dependent (BOLD) contrast, which indirectly detects neuronal activity by imaging the accompanying increase in blood flow to the local vasculature. The increased blood flow overcompensates for increased oxygen consumption, such that the ratio of oxyhaemoglobin to deoxyhaemoglobin rises in the active brain region. Deoxyhaemoglobin is often referred to as a natural contrast agent: as a paramagnetic substance, it has a shorter $T_1^*$ than oxyhaemoglobin (which is not paramagnetic, but diamagnetic). Thus, it is the changes in the deoxyhaemoglobin level which are imaged in fMRI.

**Brain Activation:** During an fMRI study, subjects are imaged whilst simultaneously being presented with specific tasks or sensory stimulations which produce activity in certain regions of their brain. Common examples of stimuli/tasks include

1. Listening to sound tones for activation of the auditory cortex
2. Observing flashing lights (photic stimulation) for activation of the visual cortex
3. Finger/thumb motion for activation of the sensory motor cortex

**Scanning Parameters:** Since deoxyhaemoglobin has a shorter $T_1^*$ than oxyhaemoglobin, all fMRI images are $T_1^*$ weighted. As gradient echo EPI (GE EPI) has a strong $T_2^*$ weighting and rapid acquisition times, it is an ideal pulse sequence for fMRI although other techniques such as spiral imaging, conventional gradient echo imaging and FSE imaging have also been used. Static field strength is another important parameter: the BOLD contrast increases with increasing field strength. For this reason, higher field strengths are often used for fMRI (Figure F.70).

**Related Articles:** Blood oxygenation level–dependent contrast (BOLD), Oxyhaemoglobin, Brain activation


**Functional MRI (fMRI)**

(Magnetic Resonance) See Functional magnetic resonance imaging (fMRI)

**Fuse**

(General) Fuse is a safety device used to protect an electric circuit against excessive current. This is achieved by opening the electronic circuit when current exceeds the amount, determined by the rating of the fuse. Opening a circuit under high current conditions prevents equipment damages and overheating, which could cause a fire. A fuse is connected in series with the circuit and consists basically of a thin wire of low-melting alloy which is heated by the electric current passing through it due to its electrical resistance. When the current exceeds the safe value for which the fuse is designed, the wire either melts or vaporises, thus opening the circuit and stopping the current. The thin wire may be made of aluminium, tin-coated copper or nickel. The fuse housing in electronic equipment is most often a cylindrical glass or ceramic type with a metal cap at each end. It is designed to resist the pressure generated by the wire vaporisation, provided the voltage across the fuse does not exceed its rating. There are two basic types of fuses: fast acting and slow blow. The fast-acting fuses blow very quickly when their particular current rating is exceeded. Slow-blow fuses have a coiled construction inside. They are designed to open only on a continued overload, such as a short circuit. The purpose of coiled construction is to prevent the fuse from blowing on just a temporary current surge. A slow-blow fuse is usually used to protect motors, and a fast-blow fuse to protect electronic equipment.

**FWHM (full width at half maximum)**

(Nuclear Medicine) See Full width at half maximum (FWHM)

**FWTM (full width at one-tenth maximum)**

(Nuclear Medicine) See Full width at one-tenth maximum (FWTM)
Gadolinium chelate

_Magnetic Resonance_ Gadolinium chelates are by far the most commonly used form of contrast agent in MRI. They are a form of paramagnetic contrast agent, usually used as a positive, $T_1$-shortening agent. The dominant position of gadolinium is attributable to its large relaxivity (having seven unpaired electrons) and large number of coordination sites (nine).

A gadolinium chelate consists of a $\text{Gd}^{3+}$ ion combined with a chelate molecule, the purpose of which is to bind the ion which is highly toxic in its free form, causing rapid liver necrosis. A number of agents are commercially available, each based on a different chelate molecule (Figure G.1).

Gadopentetate dimeglumine was the first agent to become commercially available. The metal-chelate complex carries a net negative charge, requiring neutralisation by the addition of meglumine (an organic molecule containing ammonium, $\text{NH}_3^+$). A class of nonionic agents was later developed (e.g. gadodiamide) in which the metal-chelate complex is uncharged. These agents have much lower osmolarity and from this perspective are arguably more suitable for in vivo use.

Gadolinium chelate agents are usually administered by intravenous injection, travelling in the circulation and freely crossing into the intravascular and intracellular spaces. Enhancement is therefore noted in areas of enhanced vascularity, such as angiogenesis associated with a tumour. They do not cross the intact blood–brain barrier, leading to prominent enhancement in the presence of a lesion causing disruption of this barrier (Figure G.2).

These agents have a range of clinical applications, including characterisation of tumours, visualisation and dating of cerebral ischaemia, and assessment of areas of inflammation and infection. There is a growing area of clinical applications in cardiac imaging, including assessment of myocardial perfusion and of viability following infarction.

The superior safety profile of gadolinium chelates relative to x-ray contrast agents, which can be nephrotoxic, combined with the increasing range of clinical indications for MRI, led to widespread use of these agents, including in patients with renal problems. In some cases, such as for MR angiographic studies, multiple doses of contrast agent were used. In 2006, an association was found between administration of gadolinium-based contrast agents and a condition known as nephrogenic systemic fibrosis (NSF) in patients with severe renal impairment. The detailed mechanism remains under investigation at the time of writing (2009), but putatively the increased biological half-life of gadolinium agents in the case of renal dysfunction increases time for release of free gadolinium, which may be the trigger for NSF. Theoretically, ionic cyclic agents would be expected to be the most stable and nonionic linear agents the least stable, and this appears to match the pattern of NSF incidence. Careful screening for renal dysfunction, use of more stable agents and less use of multiple doses have virtually eliminated new cases of NSF.

Related Articles: Contrast agent, Paramagnetic contrast agents, Positive contrast media

Gadolinium orthosilicate (GSO)

_Nuclear Medicine_ GSO is an inorganic scintillation crystal which is used due to its relatively high atomic number ($Z = 64$). The crystal can be grown into reasonably large sizes, which is another attractive feature. The light yield and the decay time of GSO depend on the level of doping, i.e. the amount of impurities added to the crystals. The maximum light yield is observed when the doping concentration of cerium is about 0.5 mol%. The decay time is dominated...
Gadolinium oxysulphide (Nuclear Medicine) Gadolinium oxysulphide (Gd₂O₂S) is an inorganic scintillator. When used in medical imaging it is doped with terbium. Gadolinium oxysulphate is commonly abbreviated as GOS. The wavelengths emitted in the scintillation process range from 382 to 622 nm with a primary emission peak at 545 nm.

Related Article: Inorganic scintillators

Gafchromic film (Radiotherapy) Gafchromic film is the popular vendor name of the self-developing radiochromic film (i.e. a film undergoing a change in colour corresponding to exposure to radiation) used for film dosimetry in various areas of medical physics (radiotherapy, diagnostic radiology, etc.). The sensitivity and the thickness of the emulsion vary depending on the use.

The most common Gafchromic film is a light blue coloured film, has an almost tissue-equivalent composition and develops a darker blue colour when exposed to radiation. Other films are also available such as a yellow coloured film which develops a black colour when exposed to radiation. This colouration results from the polymerisation of a dye. The polymer will absorb light and so by using a densitometer is it possible to measure the transmission of light through the film and relate this to the level of radiation exposure.

With careful handling and calibration of the film it should be possible to achieve a precision of better than 3% for dosimetric work. The film is becoming increasingly popular in general use due to the loss of film processing facilities in hospitals and there has also been a reduction in the cost of radiochromic films. Gafchromic film is commonly used in brachytherapy and stereotactic radiotherapy due to its high spatial resolution and response for high doses, as well as for IMRT verification and general linac QA measurements.

Abbreviations: IMRT = Intensity modulated radiation therapy and QA = Quality assurance.

Related Article: Radiochromic film

Hyperlink: www.gafchromic.com

Gain (Nuclear Medicine) The gain of electrical/electronic circuits is defined as the ratio between the output and the input signal.

When talking about gain in nuclear medicine one often refers to the photomultiplier (PM) tube signal gain. A photomultiplier tube has a gain up to 10⁸ depending on the design of the PM tube.

Related Article: Photomultiplier (PM) tubes

Gain (Ultrasound) In ultrasound imaging, gain is a control of the amplification of ultrasound signals. Gain can be applied to B-mode, M-mode, colour flow and spectral Doppler signals. It is an important control in everyday operation of scanners. The effects in ultrasound scanners are:

B-mode: Increased gain makes the overall image appear brighter, conversely lowering the gain reduces all the levels used in the image (Figure G.3). The gain control modifies grey levels throughout the image, this can be modified by the depth/time gain.

Figure G.2 Rim enhancement of a tumour disrupting the blood–brain barrier.

Figure G.3 Image of a kidney (top) and (lower) the same view with the gain increased by 17 dB. With high gain the whole image appears brighter but dark grey levels are not used in the image and contrast is reduced and detail lost.
Gain, amplification

Gain correction

(Nuclear Medicine) Gain correction refers to compensations for non-linearities in electric charge of the multiplier response.

The goal of gain correction is that every pixel should yield identical output signals for a given input signal and that this relationship is true for all input intensities. The gain can also differ between two individual photomultiplier (PM) tubes. Such unwanted difference in gain is corrected by a uniformity correction.

Abbreviation: PM tubes = Photomultiplier tubes.

Related Articles: Photomultiplier (PM) tubes, Uniformity

Gain factor

(Nuclear Medicine) The factor by which an electrical signal is magnified in an electronic charge multiplier, e.g. a photomultiplier tube.

The gain factor in a photomultiplier (PM) tube used for nuclear medicine imaging depends on PM tube design, but can be as high as 10^6.

Related Article: Photomultiplier (PM) tubes

Gallium-67 [67Ga]

(Nuclear Medicine)

Element: gallium (group III-A)
Isotopes: 56 < N < 86
Atomic number (Z): 31
Neutron number (N): 36
Symbol: 67Ga
Production: Cyclotron, e.g. 66Cu(α,2n)67Ga → 67Zn
Daughter: 67Zn
Half-life: 3.26 days
Decay mode: EC-decay
Radiation: gamma, internal conversion electrons, Auger electrons, characteristic x-ray photons
Gamma energy: 93.31 keV (39.2%), 184.58 (21.2%), 300.21 keV (16.8%), 393.53 keV (4.7%)
Dose rate from 1 MBq: 0.0208 μSv/h at 1 m; 208 μSv/h at 1 cm
Absorption (HVL): 1 mm lead
Biological half-life: 12 years (citrate)
Critical organ: bone surfaces, lower large intestine, red bone marrow, spleen, adrenals.
ALImin (50 mSv): 300 MBq
Absorbed dose (gallium citrate): 0.59 mGy/MBq bone surfaces, 0.20 mGy/MBq lower large intestine, 0.19 mGy/MBq red bone marrow, 0.15 mGy/MBq spleen, 0.14 mGy/MBq adrenals.
Effective dose: 0.12 mSv/MBq

Clinical Applications: In contrary to indium, which is labelled to a number of pharmaceuticals, the only existing radiopharmaceutical made by gallium is 67Ga-citrate. It is available as a sterile aqueous solution, in most cases with an activity of 37 or 74 MBq in 1 mL. 67Ga-citrate is prepared by neutralising acidic NCA 67Ga-chloride with sodium hydroxide with 4% sodium citrate. The injection solution has pH 5.5–8. The recommended administered activity is 74–185 MBq.

67Ga-citrate can be used clinically for non-specific imaging and/or localisation of non-Hodgkin’s disease, lymphoma, and bronchogenic carcinoma. 67Ga-citrate is also used for the localisation of inflammatory lesions (e.g. sarcoidosis) or fever of unknown origin.

Related Articles: Indium-111, Gallium-68

Gallium-68 \[^{68}\text{Ga}\]  

**Gallium-68 \[^{68}\text{Ga}\]**  
**(Nuclear Medicine)**  
Element: gallium (group III-A)  
Isotopes: 56 < N < 86  
Atomic number (Z): 31  
Neutron number (N): 37  
Symbol: \(^{68}\text{Ga}\)  
Production: Generator \(^{68}\text{Zn}(p,n)^{68}\text{Ga}\) → \(^{68}\text{Zn}\)  
Daughter: \(^{68}\text{Zn}\)  
Half-life: 67.63 min  
Decay mode: \(^{\beta^+}\) (89%) and EC (11%)  
Radiation: \(^{\beta^+}\) (max 1899 keV), annihilation photons  
Photon energy: 511 keV (178%)  
Dose rate from 1 MBq: 0.140 μSv/h at 1 m; 1400 μSv/h at 1 cm  
Absorption (average range of \(^{\beta^+}\)): 2.15 mm in tissue  
Critical organ: small intestine, upper large intestine, red bone marrow, bone surfaces, spleen and adrenals  
ALL\(_{ad}\) (50 mSv): 600 MBq  
Absorbed dose (gallium citrate): 0.064 mGy/MBq small intestine, 0.053 mGy/MBq large intestine, 0.046 mGy/MBq red bone marrow, 0.037 mGy/MBq bone surfaces, 0.036 mGy/MBq spleen, 0.034 mGy/MBq adrenals.  
Effective dose: 0.027 mSv/MBq

Clinical Applications: Its clinical use is so far limited, but theoretically the same as for the gallium isotope \(^{67}\text{Ga}\), i.e. for non-specific imaging and/or localisation of non-Hodgkin’s disease, lymphoma, bronchogenic carcinoma, and inflammatory lesions (e.g. sarcoidosis) or fever of unknown origin using PET.  
Because of a radioactive equilibrium between \(^{68}\text{Ge}\) and \(^{68}\text{Ga}\), the first-mentioned nuclide is used as standard calibration source for PET cameras.

**Related Article:** Gallium-67


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**Gamma camera**  
(*Nuclear Medicine*) See Scintillation camera

**Gamma camera SPECT systems**  
(*Nuclear Medicine*) See Single photon emission computed tomography (SPECT)

**Gamma correction tables**  
(*Diagnostic Radiology*) Gamma correction tables are rarely used today. These are necessary in radiography when the type of x-ray film is changed (a new film with different gamma or latitude is used). The tables show the necessary change of some of the basic radiographic parameters (kV, mA s) which will present over the new film the usual anatomical structures with the same optical densities (gray levels). The tables are made for each anatomical structure and each x-ray equipment in the department. The tables are usually made by the respective senior radiographer and medical physicist. In digital radiography similar adjustment (presenting identical pixel values for identical structures, while x-raying with the same radiographic parameters) is usually made by the service engineers during the system maintenance.

**Related Articles:** Exposure point, Film latitude

**Gamma knife**  
(*Radiotherapy*) The gamma knife is a radiotherapy device for treating brain tumours using external-body radioactive sources originally developed by Leksell. A shielded applicator is placed around the patient’s head. Up to 201 cobalt 60 sources are placed inside the applicator with the shielding used to collimate the dose to deliver a high intensity to a target point in the patient’s brain. The gamma knife is used to deliver radiosurgery, i.e. a single fraction treatment. The applicator is affixed to the patient’s skull to ensure it remains fixed relative to the treatment target. This contains the collimation system for the sources.

**Related Articles:** Stereotactic, Radiosurgery, Stereotactic frame

**Gamma radiation**  
(*General*) Gamma radiation is one type of electromagnetic radiation, characterised by its range of energy and its mode of production. It has no upper limit to its energy and the lower limit is approximately 0.1 MeV. It is produced as a result of nuclear transformations. At the lower end of its energy range it overlaps, in terms of energy, with x-rays.  
Gamma radiation and gamma rays are synonymous terms.

**Related Article:** Nuclear transformations

**Gantry**  
(*Radiotherapy*) The gantry is the part of the radiotherapy or imaging equipment supporting the radiation source and the collimation system and allowing their movements. The gantry can be moved in a circular path permitting the sources of radiation to rotate on about a horizontal axis (gantry axis). During the rotation of the gantry the beam axis and the collimator axis move in a vertical plane around a point called the isocentre.  
The gantry (Figure G.5) is the part of the radiotherapy or imaging equipment supporting the radiation source and the collimation system and allowing their movements. The gantry can be moved in a circular path permitting the sources of radiation to rotate on about a horizontal axis (gantry axis). During the rotation of the gantry the beam axis and the collimator axis move in a vertical plane around a point called the isocentre.
Gas amplification

The gas amplification in gas-filled radiation detectors depends not only on the kind of gas and its pressure but also on the voltage between electrodes. In Figure G.6 a scheme of the measured current versus potential between the electrodes for x- or gamma radiation is shown.

The gas amplification occurs when the primary electrons (e.g., photoelectrons, Compton electrons produced by x- or gamma radiation) and ions accelerated by the potential between the electrodes gain sufficient energy to ionise the gas and produce secondary electrons that can create further ionisation. The electric charge \( Q \) created by \( n_0 \) primary electrons and ions is equal:

\[
Q = n_0 e M
\]

where

- \( e = 1.6 \times 10^{-19} \text{C} \)
- \( M \) is the average gas amplification factor

The value of gas amplification is proportional to the applied voltage and depends on the detector dimensions.

**Related Article:** Proportional counter


Gas flow counters

Gas flow counter are used for small samples emitting alpha and beta particles or soft photon radiation.

The radioactive sample is introduced in the gas-filled detector (Figure G.7) operating in the proportional region (see the following Related Articles). The gas flows through the detector during the counting. The counting efficiency is almost 100%.

Usually noble gases in binary mixtures are used (e.g., a mixture of 90% argon and 10% methane) to ensure good value of the gas multiplication factor.

If the aim of the studies is an evaluation of the deposition of ionising radiation energy in biological tissue, it is necessary to use a gas mixture consisting of 64.4% methane, 32.4% carbon dioxide and 3.2% nitrogen.

The same technique can be used in flow Geiger–Müller counters. In a flow Geiger–Müller detector the sample, if it is in the form of gas, can be mixed with the counter gas (argon-alcohol). This method is applied to detect low concentration of C-14 introduced in the form of CO₂ and H-3 added as a vapour of the tritiated water.

**Related Articles:** Gas-filled radiation detectors, Geiger–Müller (GM) counters, Proportional counter


Gas-filled radiation detectors

In gas-filled radiation detectors the ionised gas molecules and electrons are collected by electrodes (Figure G.8) as current or charge due to the applied voltage. The quantity of current or charge depends not only on the kind of gas and its pressure but also on the voltage between electrodes. In Figure G.6 the variation of the measured current versus potential between the electrodes for x- or gamma radiation is shown.

Photons (x or gamma) passing through gas can interact with its molecules by the photoelectric or Compton effects. If their energy...
is equal or higher than 1.022 MeV they can produce electron-positron pairs. The electrons resulting from these processes are accelerated by the applied voltage between the electrodes and have enough energy to ionise the gas. In ionisation chamber region (1) all electrons and ions are collected by the electrodes. The measured current (charge) is proportional to the initial energy of radiation. If the applied voltage increases the electrons and ions acquire increased kinetic energy and can produce additional ionisation which gives a multiplication of collected charge (multiplication in range $10^2$–$10^3$).

This is the proportional counter region (2), the current is proportional to the initial quantity of radiation (initial number of ions). At higher voltage is the region of limited proportionality. Then we pass to the Geiger region (3) where an avalanche of interactions is generated and relatively large gas amplification occurs. In a Geiger region of operation the measured signal (current or charge) is independent of the initial energy of radiation. The number of ion pairs produced in one discharge is about $10^9$–$10^{10}$ and the amplitude of signals is so large that (its) amplification is not used. Further increase of the voltage produces spontaneous discharge which may damage the detector. This is the (a) continuous discharge region (4).

Related Articles: Geiger–Müller (GM) counters, Ionisation chamber, Proportional counter


Gate

(Ultrasound) Gate is a term used in pulsed wave Doppler denoting the range over which Doppler investigation is conducted along the Doppler beam. Gate is also described as range gate or sample volume.

See Sample volume

Related Articles: Sample volume, Pulsed wave Doppler

Gated SPECT

(Nuclear Medicine) Gated SPECT imaging makes it possible for a temporal display of the perfusion distribution of the myocardium during a heart cycle. Gated myocardial perfusion SPECT imaging is an acquisition synchronised to the patient’s electrocardiogram (ECG). A number of projection images (8 or 16, also called intervals or frames) are acquired at each projection angle, with each image or frame corresponding to a specific part of the cardiac cycle. For maintaining adequate count density of the individual cardiac frames a high count density, often $^{99m}$Tc-labelled tracer is used. All projection images for a given cardiac interval are reconstructed separately to a SPECT image volume. The SPECT volume corresponding to the various cardiac intervals can be viewed in 4D format allowing for evaluation of dynamic cardiac function.

The determination of the myocardial mass, the endocardial, the epicardial surface and the valve plane for each gating frame allows for calculation of the end-diastole and the end-systole volume from which the ejection fraction can be derived. Gated SPECT images thus allows for evaluation of both perfusion and function of the myocardium. The extent and severity of the perfusion defect, left ventricular ejection fraction as well as regional and global ventricular functions can be assessed.

Related Articles: Severity, Extent

Gating, cardiac

(Magnetic Resonance) See Cardiac gating

Gating, respiratory

(Radiotherapy)

Imaging Gating: Gating may be used in radiological imaging to reduce the artefacts due to the effects of organ motion during the cardiac and respiratory cycles. In imaging for radiotherapy, the respiratory cycle is particularly problematic as a treatment irradiation typically lasts one or several minutes, encompassing several breathing cycles, during which internal anatomy, including the tumour, may move significantly. Gating may be used to acquire a snapshot scan at a particular phase of the cardiac or breathing cycle.

An emerging use of gating is so-called 4D acquisition, particularly in CT. A scan is taken of the patient over several breathing cycles and each data point is labelled with phase of the cycle. The data are then reconstructed to yield several scans covering several phases throughout the cycle.

Treatment Gating: Gating of treatment is used in external beam radiotherapy to minimise the effects of motion, specifically breathing motion, on the treatment accuracy. This technique makes it possible for the clinician to define a smaller target volume so as to protect the organ at risk (the lung) and increase dos to the target volume. Two essentially different approaches exist: to gate the patient’s breathing using active breathing control (ABC) or deep inspiration breath hold (DIBH); to measure the patient’s breathing and to gate the beam delivery by turning it on and off depending on the phase and/or position in the breathing cycle. Issues associated with gating the treatment include the duty cycle (i.e. the fraction of time the radiation beam is being delivered) and the characteristics of the beam as it is repeatedly turned on and off by the gating process. A variety of methods exist to measure the position in the breathing cycle, including the use of one (or several) reflective markers on the patient’s surface, a pressure sensitive belt, spirometry and x-ray imaging. (an other approach is developed by a company using fiducial markers implanted near the diaphragm. These markers are tracked by x-ray imaging – CF Brainlab).

Abbreviations: ABC = Active breathing control, CT = Computed tomography and DIBH = Deep inspiration breath hold.

Related Articles: Respiratory gating, Active breathing control, Deep inspiration breath hold technique

Gauss

(General) Gauss (G) is a c.g.s unit for magnetic flux density B (also magnetic induction):

$$1 G = 10^{-4} \text{T}.$$
Gaussian distribution

(c.g.s. – the centimetre, gram second system of units, the c.g.s is now replaced by SI).

One gauss is defined as one maxwell per square centimetre.

In the SI system magnetic flux is measured in weber (Wb), thus magnetic flux density of 1 T is

\[ 1 \text{T} = 1 \text{Wb/m}^2 \] (weber per square metre)

The gauss is named after the German physicist (and mathematician) Karl Friedrich Gauss.

Related Article: Tesla

Gaussian distribution

(General) The Gaussian distribution, or the normal distribution, is a continuous probability distribution that has a mean value and a variance. The standard Gaussian has a mean of zero and a variance of one. The shape of the distribution looks very much like a bell that falls rapidly towards plus/minus infinity.

The equation for the Gaussian distribution is given by

\[
\frac{1}{\sigma \sqrt{2\pi}} \exp \left( -\frac{(x-\mu)^2}{2\sigma^2} \right)
\]

where

- \( \mu \) is the mean
- \( \sigma \) is the standard deviation (variance)

Gaussian functions (or filters) are widely used in image processing to characterise the blurring effect of scintillation camera collimators.

Gaussian noise

(General) Gaussian noise is noise whose probability distribution is that of the Gaussian distribution (normal distribution). Gaussian noise is equal to the standard deviation, \( \sigma \), in the Gaussian distribution equation.

Related Articles: Gaussian distribution, Poisson noise, Poisson distribution

GE

(Magnetic Resonance) See Gradient echo (GE)

Geiger–Müller (GM) counters

(Radiation Protection) The Geiger–Müller counter (G–M counter or Geiger tube) was invented in 1928 by Hans Geiger and Walther Müller. It is a gas-filled detector based on ionisation. The construction is similar to ionisation chambers and proportional counters but it works on the principle of gas multiplication. As a result all pulses from a G–M counter are of the same amplitude whatever the energy of the incident radiation. A diagram of a G–M counter is shown in Figure G.9. The hollow metal tube (or glass tube with mesh cylinder) is negatively charged and filled with argon or neon gas at low pressure. There is a positively charged electrode along the centre of the tube. X-ray, gamma photons or other particles (if the end-window is sufficiently thin) entering the counter cause ionisation of the gas. The potential difference between electrodes is several hundred volts. Electrons released as a result of the ionisation are attracted to the positively-charged central anode. As they pass through the gas they are accelerated, gaining sufficient energy to eject electrons from other gas atoms. These new electrons are also accelerated, and may cause further electrons to be released – a chain reaction. One gamma photon can produce about 10^5 electrons. Quenching of the gas multiplication process is necessary to limit the amount of detector dead time. The quenching is accomplished with use of a small quantity of alcohol vapour in the gas.

A graph of the operating characteristics of a gas counter is shown in Figure G.10. The region of the graph that a G–M counter operates in is region b, the plateau region corresponding to the maximum gas amplification. The plateau is between 100 and 1500 V. It depends on the size of the G–M counter.

In Figure G.11 the scheme of the G–M counter electronic system is shown.

Related Articles: Gaussian distribution, Poisson noise, Poisson distribution


Gel

(General) A gel is a crosslinked substance which exhibits thixotropy, i.e. it is a solid at rest and becomes fluid when agitated. A gel is mostly fluid, but behaves as a solid due to the networked structure
within the liquid. The density of a gel is similar to its constituent liquid. The crosslinking contributes to a gel’s hardness and toughness, which can vary considerably. Many fluids can be used to form a gel including water (hydrogels), oil (organogels) and air (aerogels).

Gels are extremely versatile in their uses, ranging from foods to adhesives. They are used for fibre optic communication and to prevent water intrusion and mechanical damage. A cationic polymer is often used in gels because its positive charges prevent coiling, allowing the polymer to increase the gel’s viscosity. These polymers are used in hair gels as they bind to the anionic amino acids on the keratin molecules in hair.

Hydrogels contain natural or synthetic water-insoluble polymers with hydrophilic groups, for example acrylates and agarose. They are very absorbent and flexible, similar to biological tissue, due to their substantial water content. They can be made to be sensitive to changes of pH, temperature, or concentration of a metabolite, causing them to swell as a result. Hydrogels are also used in disposable diapers (nappies) to absorb urine.

Organogels are non-crystalline thermoplastic solids in organic liquids such as vegetable oils. The solubility and particle size determine the elastic properties and hardness of an organogel. Organogels are used in pharmaceuticals, cosmetics, art conservation and food.

Xerogels are solids formed from gels by drying them with unconstrained shrinkage. They maintain a very low density, a high porosity and a large surface area. Formation under hypercritical conditions minimises shrinkage and decreases the gel’s density to produce an aerogel. Conversely, heat treatment causes viscous sintering, thereby maximising shrinkage to produce a dense glass. An example is silica gel used in chromatography and for thermal insulation.

Medical Applications: Hydrogels are extensively applied in medicine for purposes such as topical drug delivery, burn dressings, medical electrodes and breast implants. They can be used as tissue phantoms in medical physics because of their tissue-like properties. They are also used as an ultrasound gel to acoustically couple the transducer to the skin surface. More advanced uses include radio-sensitive gels for dosimetry, biosensors responsive to specific molecules, and in tissue engineering in the form of scaffolds containing human cells to repair tissue.

Related Articles: Adhesives, Fricke based gel, Gel dosimetry, Polymer gels, Ultrasound

Gel dosimetry
(Radiotherapy) Absorbed dose measurement using nuclear magnetic resonance (MR) was first introduced by Gore et al. (1984), who proposed that the changes induced by the interactions of ionising radiation in the Fricke ferrous sulphate dosimeter solution could be probed with MR rather than using the different light absorption properties that can be quantified by UV spectrometry. Using the MR specific relaxation time parameters $T_1$ and $T_2$, it is possible to measure the radiation dose of the well-known Fricke dosimeter. Later the Fricke solution was incorporated in a gel matrix thus realising a direct 3D dosimeter. The purpose of the gel matrix is to prohibit radiation product movement from the place of formation to surrounding regions thus saving the record of 3D spatial dose distribution. The number of ferric ions being produced have been shown to be increased from approximately 15 Fe$^{3+}$ ions per 100eV to more than 100 Fe$^{3+}$ ions per 100eV. Organic additives, such as benzoic acid, enhance the Fricke dosimeter chemical yield reacting with O$_2$. The resulting peroxides oxidise ferrous ions into ferric ions thus triggering the increase of the dose-response sensitivity. In practice the ferric ions produced by the absorption of radiation diffuse readily through the gel or agarose matrix, leading to a decrease in signal intensity, and a loss of spatial information. Therefore MR imaging of the irradiated gel must be performed within few hours of irradiation to avoid serious degradation of the dosimetric detail. A unique and important characteristic of the gel dosimetry is that the measurement is totally non-invasive. The gel is a tissue equivalent material and there is the coincidence of the irradiated medium and dosimeter which can be moulded into arbitrary shapes. Therefore there is no necessity to place a probe into a phantom thus causing a perturbation of the radiation fluence. It is also not necessary to remove part of the irradiated material to perform the measurement. The radiation chemistry of the dosimetry system and the MR processes are well understood. Another specific characteristic of the gel dosimetry is that the point of measurement in gel is determined completely by the measuring system which can be programmed to scan the full volume to obtain a complete 3D dose distribution. The dose integration in the gel dosimeter permits measurements of dynamic and multiple beam treatments. The strong limitation of the Fricke infused gel dosimetry is the post irradiation diffusion of the ferric ions which tends to blur the image altering the determination of the dose distribution over time thus demanding that the MR reading of the dosimeter take place as soon as possible after irradiation.

The accuracy and the sensitivity of an individual gel batch depend on the preparation conditions and the chemical purity of components; therefore it is recommended to perform a separate calibration of the batch at the time of use. Several methods of gel dosimeter calibration have been reported but in many of them a quantity of the gel batch is transferred to a calibration phantom and then irradiated at a series of known doses. To obtain the calibration plot $R_2 = 1/T_2$ values must be determined for each incremental dose. This is achieved by measuring the signal intensity within each flask at each echo time and fitting data by

$$\ln S(TE) = \ln S_0 - R_2(D)TE$$

where

$S(TE)$ is the measured signal intensity at a given echo time $TE$
$S_0$ is the signal at $TE = 0$
$R_2(D)$ is the transverse relaxation rate being function of the dose $D$

A calibration is then established by plotting the estimated $R_2$ for each known dose value which should fit the linear equation

$$R_2(D) = \frac{\Delta R_2}{\Delta D} D + R_2^0$$

where

$R_2^0$ is the intercept
$\Delta R_2/\Delta D$ is the slope of the fitted line to the calibration data

Figure G.12 shows values of transverse relaxation rates ($R_2$) for specific gels as a function of dose. The data show that the dose response is well fitted by a straight line and is highly reproducible over a wide range of doses. An indication of the sensitivity of the gel dosimeter can be gauged by the ratio of the slope to the intercept.

Related Article: Polymer gel dosimetry


**General public exposure**

*(Radiation Protection)* General public exposures are incurred by members of the public from radiation sources, excluding any occupational or medical exposures and the normal local background radiation, but including exposures from authorised sources and practices and from intervention situations.

Also known simply as public exposures, they have defined dose limits.

**Related Articles:** Public exposures, Dose limits


**Generator**

*(General)* See AC generator

**Generator, battery powered**

*(Diagnostic Radiology)* This x-ray generator is usually used in low power mobile x-ray equipment. The principal diagram of a battery-powered x-ray generator is given on Figure G.13. The DC voltage from the battery (accumulator) is converted to pulses by a DC/AC converter (inverter or chopper). The pulses are usually with frequency 0.5–2 kHz. These present AC voltage, which supplies the high voltage transformer and produces the x-ray tube anode voltage (kV). Due to the higher frequency a ferrite-core transformer can be used, what virtually transfers this generator to a type of medium (high) frequency generator.

**Generator(s), three-phase**

*(Diagnostic Radiology)* See Three-phase generator

**Generator, capacitor-discharge**

*(Diagnostic Radiology)* See Capacitor-discharge generator

**Generator, falling load**

*(Diagnostic Radiology)* See Falling-load generator

**Generator, high-frequency**

*(Diagnostic Radiology)* See High-frequency generator

**Generator, falling load**

*(Diagnostic Radiology)* See Falling-load generator

**Generator, high-frequency**

*(Diagnostic Radiology)* See High-frequency generator

**Generator kV waveform**

*(Diagnostic Radiology)* See Voltage waveform

**Generator, single-phase**

*(Diagnostic Radiology)* See Single phase generator

**Generators, radionuclide**

*(Nuclear Medicine)* See Radionuclide generators

**Genetically significant dose (GSD)**

*(Radiation Protection)* The genetically significant dose (GSD) is an indication of the overall genetic risk. Typically the human beings are exposed to different kind of ionising radiation exposures: natural background, occupational, consumer products, environmental sources, nuclear power and medical exposures (diagnosis and therapy). The natural background which includes radon (the largest and most variable component), cosmic rays, internal radioactivity and terrestrial radioactivity, is by far the biggest source of radiation. Man-made radiation includes medical exposures and consumer products. The genetically significant dose can be estimated for populations, groups of population (as example occupational workers, who should pay particular attention) and also deriving from particular items to a population (as, e.g. was done for luminous wrist watches, etc.).

**Hyperlinks:** [https://www.NCRP.org](https://www.NCRP.org); [https://www.IAEA.org](https://www.IAEA.org)

**Geometric distortion**

*(Magnetic Resonance)* Geometric distortion is the deviation of points in an image from their true position in an object. This artifact appears as either displacement of a point or improper scaling and is primarily caused by a non-uniform main magnetic field or non-linear magnetic field gradients.

Geometric distortion is commonly assessed through the measurements of distances on images phantoms containing straight lines or arrays of points with known lengths. An example is the
Geometric distortion
(Ultrasound) A certain amount of geometric distortion is always present in ultrasound imaging. This is inherent in the technique as the time for an echo to return is assumed to be proportional to distance. At the same time, the presence of echoes is dependent on mismatches in speed of sound. In the body, an average speed of sound of 1540 m/s is assumed. Fat on the other hand has a speed of sound which is around 1430 m/s. Mismatches in speed of sound can also lead to degraded image quality: a focussed beam will defocus through a fat layer, leading to less sharp ultrasound images.

Geometric efficiency
(Diagnostic Radiology) In CT scanning, geometric efficiency refers to the geometric dose efficiency of the detector array. In general terms, it is the percentage of photons exiting the patient that fall on the active detector area. Geometric efficiency is reduced by inactive areas within the detector array, such as the septa between detector elements, or if the x-ray beam is greater than the active detector area.

More specifically, the term geometric efficiency (GE) is used when referring to the geometric dose efficiency along the z-axis or the central axis of the beams.

Geometric error
(Magnetic Resonance) Geometric distortion can be described as the positive distortion (geometric error) in the x, y and z planes of the image. Geometric error is a useful way of quantifying geometric distortion and can be used to monitor the systems geometric distortion over time and also compare it with other systems. For a modern MR system a maximum geometric error on the order of 1–2 mm in any direction would be expected.

The geometric error can be measured simply by making length measurements from readily identified locations on an image in the three different planes and comparing them with the known values for those lengths.

Geometric field separation
(Radiotherapy) Sometimes the area to be treated cannot be covered by one field. If this is the case, then two (or more) fields may be applied. Because radiation fields diverge it is important to know the point where the geometric edges meet or overlap in order to provide an effective treatment and to avoid over-radiation or under-radiation. Calculating the separation of fields can be done with respect to the geometric edges of the beam (see also Geometric field size).
Figure G.17 shows two beams set to overlap at a depth d below the surface of the patient. The overlap point can of course be set to any convenient depth by altering the central axis separation or the geometric field edge separation.

**Related Article:** Geometric field size

**Geometric field size** *(Radiotherapy)* The shape and size of the radiation field is affected by the size of the source and the effect of scatter. In the case of a linear accelerator the effective source size would be the electron spot which is on the order of a few millimetres and in the case of cobalt it is around 15 or 20 mm. Figure G.18 shows the result for a typical source. The geometric field size is generally defined as the distance between the 50% isodose lines at opposite sides of the field, the maximum dose being taken as 100% (see Figure G.18). The field size is defined for the specified source – surface distance at the specified depth.

Geometric field size is usually referred to as radiation field size or radiation beam field size.

The source size together with the collimator transmission and scatter affects the width of penumbra region. The larger the source, the higher the portion of the source, which is shielded by the collimator, and the broader the resultant penumbra. Cobalt beams therefore will have a broader penumbra than linear accelerators because they have a larger effective source size.

**Related Articles:** Penumbra, Beam limiting device, Collimator, Collimation


**Geometric unsharpness** *(Diagnostic Radiology)* Geometric unsharpness is a term used in the past to refer to what is better described as focal spot blurring in x-ray imaging. The amount of focal spot blurring is related to the geometry and relative positions of the focal spot, receptor, and object within the body as illustrated on Figure G.19. Here the object

### Table: Z-axis geometric efficiency (%)

<table>
<thead>
<tr>
<th>Collimation</th>
<th>Z-axis geometric efficiency (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>5 mm (4 x 1.25)</td>
<td>67</td>
</tr>
<tr>
<td>10 mm (8 x 1.25)</td>
<td>83</td>
</tr>
<tr>
<td>20 mm (16 x 1.25)</td>
<td>97</td>
</tr>
</tbody>
</table>

**Figure G.16** Effect of z-axis x-ray beam collimation on geometric efficiency.

**Figure G.17** Geometric field separation.

**Figure G.18** Source size and scatter have an effect on the size of the penumbra of a radiation field.
(a hole in an object) is projected over the receptor/film (the white spot) together with some blur (the dark spot). The higher is the magnification (the object is far away from the film) the larger are both the white and blue spots, but their superposition leads to domination of the blur over the object image.

As the object in the patient’s body is located away from the receptor (increased geometric magnification) the blurring because of the finite size of the focal spot is increased relative to the size of the object’s image. This increases the blurring and reduces visibility of detail.

The amount of blurring relative to the size of an object is related to the size of the focal spot and the location of the object relative to the receptor and focal spot (along the ‘s’ scale) – shown in Figure G.20.

Blurring from the x-ray image receptor also has the same geometric dependence as focal spot blurring (Figure G.21).

As geometric magnification is increased, the size of an object 'as seen by the receptor' is larger. Therefore the amount of blur from the receptor and relative to the size of an object is decreased and visibility of detail is improved. This is the reason to use geometric magnification in procedures like mammography when high visibility of detail (small calcifications is required).

The amount of blurring relative to the size of an object is related to the actual blur value within the receptor and the location of the object relative to the receptor and focal spot (along the ‘s’ scale) as shown on Figure G.22.

**Germanium detector**

(Radiation Protection) Germanium (Ge) is a semiconductor with atomic number $Z = 32$. The energy needed to create an electron-hole pair is about $3\text{eV}$, much less than the $35\text{eV}$ required in gas detectors. The density of Ge is much greater than the density of gas used in gas-based detectors. Thus, in Ge detectors about 10 times more ion pairs are produced with greater probability (efficiency) than in a gas detector for a given x- or gamma-ray. The energy resolution of Ge detectors is also better than gas detectors. The atomic number of Ge is greater than that of silicon ($Z = 14$) thus Ge detectors have a larger cross-section for photon interactions and better detection efficiency than silicon detectors.

There are two types of Ge detectors: those doped with lithium (GeLi); and high-purity Ge (HPGe). The ionisation in the detector results in a small electrical current and is converted to a voltage pulse. The height of the pulse is proportional to the energy of the radiation absorbed by the detector.

The energy resolution of a semiconductor detector is defined by the full width at half maximum (FWHM) of the photopake (Figure G.23). Its value is expressed in keV and not as a percentage of the
photon energy. The energy resolution depends on the statistical fluctuation in the number of created electron-hole pairs, variation in the charge collection efficiency with detector size and electronic noise. The relative contribution of these factors depends on the energy of the x-rays or gamma radiation and on the volume of the detector. Therefore the energy resolution is specified at particular photon energies, e.g. 5.9 keV (Fe-55), 122 keV (Co-57), 662 keV (Cs-137), 1.33 MeV (Co-60). Typical FWHM values for small volume Ge detectors are 150–250 eV at 5.9 keV and 400–600 eV at 122 keV. For larger volume detectors typical values are 0.8–1.2 keV at 122 keV and 1.7–2.3 keV at 1.33 MeV.

The germanium detector has large thermal noise at room temperature which causes a high background reading. This background signal decreases as the temperature of the detector is reduced. Liquid nitrogen, at a temperature of 77 K, is used to cool the detector. Lithium drifted germanium Ge(Li) must be kept at this temperature to prevent the lithium drifting out. HPGe detectors, however, are stable at room temperature.


**Ghost artefact**

*Magnetic Resonance* A ghost artefact is a low intensity image of an object which appears in a different location from the primary signal area. When ghosting occurs there is a SNR reduction in the primary signal area. Ghosting can potentially lead to important clinical information being obscured by the ghost signal.

**Motion Ghost:** This type of ghost is caused by movement of the patient, either physiological motion such as breathing and bowel motion, or patient movement such as small twitches. The ghost appears along the phase-encoding direction. *Motion ghost* artefacts can be reduced by respiratory and cardiac triggering, the use of breath holding pulse sequences, or presaturation pulses, depending on their origin.

**Flow Ghost:** Flow artefacts are another type of motion artefact caused by the flow of blood or cerebrospinal fluid within the body. Flow compensation can be used to reduce this artefact.

**Nyquist Ghost:** The N/2 or Nyquist artefact only occurs when using an EPI pulse sequence. The artefact is caused by slight differences in alternate lines in k-space. The ghost produced is displaced half the field of view away from the original image.

**Quadrature Ghost:** RF quadrature artefacts are caused by a disturbance in the two detector channels of the quadrature detector.

**Abbreviations:** DC = Direct current, EPI = Echo planar imaging, RF = Radiofrequency and SNR = Signal to noise ratio.

**Related Articles:** Motion artefacts, N/2 artefact, Quadrature artefact

**Ghosting**

*Magnetic Resonance* Ghosting is the term used to describe a low intensity displacement of MR signal which can manifest itself either as a series of distinct low intensity copies (ghosts) of the image, or as a column of smeared signal.

The most common cause of ghosting is motion; this can be patient motion, physiological motion or blood flow. Ghosting mainly propagates along the phase encoding direction since signal sampling and spatial encoding are performed over many TRs and hundreds of milliseconds in the phase encoding direction, compared to a few milliseconds in the frequency encoding direction (Figure G.24a and b).

**FIGURE G.24** (a) Ghosting along the phase encoding direction due to eye movement. (b) Low intensity whole head ghosts (along the phase encoding direction).
Gibb's artefact

(Magnetic Resonance) See Gibb's ringing

Gibb's ringing

(Magnetic Resonance) Gibb's ringing artefacts (also known as truncation artefacts) occur near sharp boundaries and high-contrast transitions in MR images. The artefacts appear as alternate bands of high and low intensity signal, parallel to interfaces between tissue or phantom edges. Clinically, there are two main anatomical locations where Gibb's ringing occurs:

1. At the edge of the brain
2. At the interface between CSF and the spinal cord

The cause of Gibb's ringing is insufficient sampling of the high frequencies inherent at sharp edges.

Fourier Theory: To understand the phenomenon further, it is useful to consider Fourier theory, which states that any periodic waveform can be constructed from a sum of sinusoids of varying amplitude, frequency and phase.

In MRI, images are created via such a summation, where the number of frequencies present in the final image is determined by the number of phase and frequency encoding steps. For a simple example, let us consider a row of pixels through an MR image of a uniform test object (see Figure G.25). As the first (and last) voxels in this row represent regions outside the test object, they should have an intensity value of zero, whilst the central voxels (representing the test object itself) should be of constant, non-zero intensity. The off-on-off waveform representing this intensity variation is known as a 'top-hat' function (see Figure G.25).

According to Fourier theory, the 'top-hat' of voxel intensities can be approximated by summing a series of sinusoids. In Figure G.26, it can be seen the more sinusoids are summed, the better the approximation. Yet, even when summing large numbers of sines, the resultant waveform still deviates slightly from the ideal: the sharp edges of the 'top-hat' are accompanied by an unwanted oscillation of signal – this is Gibb's ringing.

In clinical MRI, Gibb's ringing is often exacerbated when the number of phase encoding steps is reduced to reduce image acquisition time. If the number of phase encoding steps is halved from 256 to 128, the acquisition time will also be halved, but so will the number of frequencies present in the final MR image. For this reason, reducing the number of phase encoding steps increases the amount of oscillatory signal (Gibb's ringing) artefact. Figure G.27a and b demonstrates this in practice.

Gibb's artefact can be avoided by increasing the number of phase encoding steps or changing the resolution of the image.

Glass envelope (of an x-ray tube)

(Diagnostic Radiology) The glass envelope provides the support of the anode and the cathode assemblies. It also keeps the deep vacuum in the x-ray tube (min 10^-4 mbar) and the high voltage insulation between the anode and cathode. The glass should have low x-ray absorption and high thermal resistance.

When multiple heavy exposures are made the anode temperature is so high, that it glows (white hot). This heat transfers to the near glass envelope. The heating and later cooling produces thermal stress, which leads to micro-cracks, what decreases the vacuum and produce arcing inside the x-ray tube (Figure G.28).

When producing the x-ray tube the glass envelope is first vacuumed and then sealed. Even so with the time the glass and the other parts in the envelope emit ions (cold emission). Special care is taken for diminishing of the cold emission – like polishing and degassing the glass and the metal electrode assemblies. However small, this emission exists and when the tube is not used for long time it can lead to arcing and damage the tube. Due to this reason x-ray tubes, which have not been used for long time must be 'degassed' – warmed slowly with low-power exposures before use (this will be explained later).

Often the glass is thinned at the exit of the x-rays (exit window) to minimise the absorption of useful radiation. Exactly at this place the metalisation of the glass is quite substantial and increases with the age of the x-ray tube (see the article on Stationary anode).

The disadvantages of the glass envelope are overcome in the metal envelope x-ray tubes.

The anode support (the shank at which the rotor with the rotating anode is coupled by the bearings) emerges from the glass envelope.

FIGURE G.25 An idealised MR image of a uniform phantom, with a corresponding 'top-hat' plot of intensity variation along a given row of pixels.

FIGURE G.26 Top-hat functions composed from varying numbers of sine waves. The initial sine waves added to the sum are of low frequency, but higher frequency sines add finer detail.
and is connected to the positive end of the high voltage. Special air-tight and heat-resistant seal is used to fix the glass envelope to the anode shank.

Related Articles: Filament circuit, Filament heating, X-ray tube, Anode, Focal spot

Hyperlinks: EMERALD (DR module), www.emerald2.eu

Gloves, lead
(Radiation Protection) See Lead gloves

Glow curve
(Radiation Protection) The glow curve is the emission spectra (Figure G.29) registered during heating of thermoluminescent material that was previously irradiated.

The glow curves of different luminescent material have well known shapes. The parameters are peak energy and full width at half maximum as well as its surface. As the light output is linear dependent on the radiation energy and dose, the evaluation of the glow curve, e.g. using an appropriate software is applied to determine the radiation dose in routine personal dosimetry practice.

Related Article: Thermoluminescent dosimeter (TLD)


Glycolysis
(Nuclear Medicine) Glycolysis is the metabolic pathway from glucose to pyruvate with a net production of two molecules of ATP per molecule of glucose. Pyruvate can then be converted to lactate or enter the tricarboxylic acid cycle. Complete oxidation of glucose through glycolysis and the tricarboxylic acid cycle yields a net production of 38 molecules of ATP, 6 molecules of CO₂ and 6 molecules of H₂O per molecule of glucose.

Glycolysis targeting
(Nuclear Medicine) This term refers to radiopharmaceuticals that target the cellular glucose metabolism. The most common glucose targeting radiopharmaceutical is ¹⁸F-FDG (2-deoxy-2-[¹⁸F]fluoro-D-glucose) which is used for clinical PET imaging. FDG is transferred into the cell by the glucl-1 transporter which is present in many cell types but overexpressed in cancer cells. Once inside the cell, the FDG undergoes phosphorylation by hexokinase which traps the FDG inside the cell. A PET imaging study will show the FDG distribution (or glucose metabolism) within the patient. This makes FDG-PET an excellent tool to detect, monitor and classify tumours.

Related Articles: Tracer kinetic modelling, Receptor targeting, Antigen targeting, DNA targeting, Neuroreceptor targeting, Apoptosis targeting, Hypoxia targeting


GMR (gradient motion rephasing)
(Magnetic Resonance) See Gradient motion rephasing (GMR)
Golay codes
(Ultrasound) A Golay coded transmission sequence is used in coded excitation schemes. The Swiss-born mathematician Marcel Golay, discovered that certain pairs of binary sequences have important correlation properties. Such pairs have equal length and the sum of their autocorrelation functions is zero except at zero lag. Examples are for instance [−1, +1], [+1, +1] and [−1, +1, −1, −1], [−1, +1, +1, +1]. Thus, range lobes are theoretically eliminated. Golay coded transmission schemes seem to be implemented successfully in different ways. One approach employs transmission of a sine burst where +1 in this sequence corresponds to a single cycle sine wave, and −1 to an inverted cycle (i.e. 180° out of phase). Another is to sequentially excite the transducer with a positive pulse, or a negative one, respectively, according to this scheme. Sufficient time is allowed between excitations to let the impulse response ring down. At least one manufacturer has employed Golay codes to improve SNR and penetration depth. Apparently the Golay coding is only used for larger depths, as the focusing at shallower regions distorts the coding to a larger extent. A drawback to using Golay codes is the need for multiple transmits, thereby reducing frame rate.


Gonad shielding
(Radiation Protection) It is a good practice to protect the gonads of young people during x-ray examinations.

For male patients there are specially shaped scrotal protections (gonad shields), in different size, usually at least four sizes, with a protection in Lead equivalent thickness of 2 mm.

For female patients there are specially shaped ovary protections in different sizes, usually five sizes, with a protection in Lead equivalent thickness of 1 mm.

For small female patients, special attention should be paid to the choice and positioning of the protection, taking into account the anatomy and the aim of the x-ray investigation. In fact a wrong choice or positioning might cover important anatomical structures and invalidate the investigation.


Good manufacturer practice
(Nuclear Medicine) Good manufacturer practice (GMP) is concerned with ensuring good quality of the production and preparation of pharmaceuticals, control and management of manufacturing and quality control testing of pharmaceutical products. A principal part of GMP is the recording and documentation of all activities during the production of the pharmaceutical enabling traceability.

Abbreviation: GMP = Good manufacturer practice.

Related Articles: Quality control, Radionuclide purity, Radiochemical purity


Gradient and spin echo (GRASE)
(Magnetic Resonance) This is a hybrid pulse sequence that combines multiple gradient and fast spin echo imaging. The pulse sequence consists of a train of refocusing RF pulses and alternating readout gradients. The refocusing RF pulses lead to a train of spin echoes, whereas the alternating gradients create multiple gradient echoes. The GRASE pulse sequence is illustrated in Figure G.30, with three readout lobes between each of the 180° refocusing pulses. GRASE images can be acquired with higher spatial resolution than those obtained from the fast spin echo sequence since more echoes per unit time can be obtained using alternating readout gradients. The SAR is also lower than for RARE because fewer RF pulses are used, allowing more slices per TR, especially at high field strength (e.g. 3.0T). Drawbacks of this pulse sequence include sensitivity to eddy currents (as known from EPI imaging), leading to phase errors. Phase corrections are necessary to remove inconsistencies between echoes originating from the alternating polarity readout.

Related Articles: Eddy currents, EPI (echo planar imaging), Fast spin echo, Specific (energy) absorption rate, SAR


Gradient coils
(Magnetic Resonance) Gradient coils generate the magnetic field gradient in the x, y and z direction (see Related Article) when current is fed through them. The gradient coils are designed in a way to create spatial magnetic field gradient as linear as possible. Furthermore, the gradient fields have to be switched on and off as fast as possible during the execution of a pulse sequence.

The hardware required to generate the gradient fields is a set of coils (electromagnets) mounted on a cylindrical former. This is most easily illustrated for the gradient coil generating the gradient field in the z-direction (the direction along the tunnel of a cylindrical magnet with a horizontal B₀-field). In Figure G.31a an example of such a gradient coil consisting of two separate windings in opposite directions (Helmholtz- or Maxwell-pair) is shown. When current (generated by the gradient amplifier) runs through these windings it results in gradient fields that either adds to or subtracts from the B₀-field. The gradient field is linear and becomes
Gradient echo (GE) is a signal generated from free induction decay (FID) by means of a bipolar switched magnetic gradient. A gradient echo sequence (Figure G.32) consists of slice selective excitation pulse to create transverse magnetisation in a slice of the object. The transverse magnetisation will decay with time (free induction decay) according to the tissue relaxation time $T_2^*$. At any time while there is still a detectable signal, a gradient echo can be obtained by applying a bipolar gradient pulse in the readout direction. When the first gradient pulse in the readout direction (here negative) is applied, spins will quickly get out of phase and no signal can be detected. When the second pulse is applied in the opposite direction the signal returns again and is at its maximum exactly when the area under the two gradient forms are equal. This corresponds to the centre of the gradient echo that is formed and is denoted the echo time (TE).

The image contrast for a gradient echo pulse imaging sequence depends not only on the echo time and the repetition time (TR) but also on the excitation flip angle ($\alpha$). Usually, a gradient echo image is measured by fast repetition of the sequence to establish a steady state of the transverse magnetisation, which is an equilibrium between the small flip angle excitation and the $T_1$-relaxation process. In general, smaller flip angles results in more $T_1^*$-weighted images and larger flip angles result in more $T_2^*$-weighted images. Specfically the following thumbs of rule apply to obtain different contrasts:

- $T_1$-weighted: Large flip angle, short TR and short TE
- $T_2^*$-weighted: Small flip angle, long TR and long TE
- PD-weighted: Small flip angle, long TR and short TE

There is also a relationship between TR, $\alpha$ and $T_c$ of the tissue, such that there is a flip angle that will result in maximum signal for a particular TR and $T_c$. It is denoted the ‘Ernst-angle’.

The lack of a refocusing (180°) pulse, which would be used to obtain a spin-echo, means that spins will lose coherence in the presence of any static magnetic field inhomogeneity. This causes severe signal loss near, for example air cavities or metal implants, but can also be very useful in, for example functional magnetic resonance imaging applications where differences in magnetic susceptibility actually results in the desired signal contrast.

Related Articles: Eddy current, Gradient field, Isocentre, Pulse sequence

Gradient linearity

Gradient linearity (Magnetic Resonance) MR systems contain three gradients for spatial encoding in the $x$-direction, $y$-direction and $z$-direction. It is important that each gradient is linear in order to obtain accurate spatial localisation. Gradient linearity is defined as the difference between the measured value of the field gradient and the ideal measure. The non-linearity is expected to be approximately 1%–2% over a 50 cm DSV. Gradient linearity tends to decrease towards the edge of the volume being imaged (Figure G.35). This results in the image being geometrically distorted at the edges of the object.

To remove some of the problems of gradient non-linearity, some manufacturers have used computer algorithms to unwarp the image, taking into account the known distortions of the real-world non-linear gradients.

**Abbreviation:** DSV = Diameter of a spherical volume.

**Related Article:** Gradient

Gradient motion rephasing (GMR) (Magnetic Resonance) In the presence of a magnetic field gradient $G$, the phase shift $\Delta \Phi$ (in the rotating frame of reference) experienced by a spin isochromat as a function of time $t$ is given by

$$\Delta \Phi(T) = \gamma \int_0^T G(t) r(t) dt,$$

where $\gamma (2\pi)^{-1} = 42.6$ MHz/T.

The gradient field is usually zero at the isocentre ($x, y$ and $z = 0$) of the magnet, so that the gradient field adds to or subtracts from $B_0$ in a linear manner as the distance from the isocentre increases. This results in a linearly varying Larmor frequency along the direction of the applied gradient field.

Figure G.33 illustrates a gradient field that is applied along the direction of slice selection while an RF-pulse is transmitted. Depending on the frequency and bandwidth of the RF-pulse the hydrogen nuclei within a slice with a specific thickness at a specific location along the slice selection direction can be excited. Note that the slice selection direction can be defined freely by the operator and the gradient field that is needed can be obtained by combining gradient fields applied in the $x$, $y$ and $z$-directions simultaneously.

The same principle is valid for the other two directions that are used for spatial encoding of an MR-image (denoted phase and frequency encoding), only that the gradient fields are applied at a different time during the execution of a pulse sequence. The frequency encoding gradient is applied during readout of the MR-signal and the phase encoding gradient is applied at some time between excitation (slice selection) and readout.

The gradient fields are switched on and off (see Figure G.34) while still avoiding a rate of change ($dB/dt$) large enough to induce currents that can cause nerve stimulation. The rise times are on the order of a few 100 $\mu$s and amplitudes are typically between 5 and 40 mT/m. The gradient fields thus are time varying magnetic fields with frequencies on the order of 1 kHz, which decrease in amplitude with increasing distance from the gradient coil.

**Related Articles:** Gradient coils, Isocentre, Pulse sequence


**FIGURE G.33** An RF-pulse is transmitted while a slice encoding gradient is applied. Only hydrogen nuclei with a Larmor frequency equal to the RF-pulse frequency are excited, i.e. those within a slice at a specific distance ($d$) from the isocentre ($d = 0$).

is switched on, the total magnetic field ($B$) that a hydrogen nucleus (proton) located at position $z$ experiences can be expressed as

$$B(z) = B_0 + G_z \cdot z [T]$$

The Larmor frequency ($f$) at that point is given by

$$f(z) = \gamma \cdot (2\pi)^{-1}(B_0 + G_z \cdot z) [Hz]$$

where $\gamma (2\pi)^{-1} = 42.6$ MHz/T.

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To remove some of the problems of gradient non-linearity, some manufacturers have used computer algorithms to unwarp the image, taking into account the known distortions of the real-world non-linear gradients.

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**Related Articles:** Gradient coils, Isocentre, Pulse sequence


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**Related Articles:** Gradient coils, Isocentre, Pulse sequence

where
\[ \gamma \] is the gyromagnetic ratio
\[ r \] is the position

The phase development of a spin is influenced by the gradient timing as well as the position of the spin at a given time, and a spin is referred to as being rephased when the net phase shift has returned to zero, i.e. \( \Delta \Phi = 0 \). Therefore, static spins are obviously rephased at time \( T \) if

\[ \int_0^T \mathcal{G}(t) dt = 0 \]

The position of an arbitrarily moving spin isochromat is given by the sum of contributions from motion components of different order, i.e. initial position (zeroth order), velocity (first order), acceleration (second order), jerk (third order), etc. When the position is described in terms of its Taylor series, i.e.

\[ F(t) = r_0 + \frac{\gamma B(t)}{2} + \frac{d\gamma B(t)}{dt} + \cdots + \frac{d^n \gamma B(t)}{dt^n} + \cdots \]

the phase shift can be expressed as

\[ \Delta \Phi(T) = \left[ \int_0^T \mathcal{G}(t) dt + \frac{\gamma B(t)}{2} + \frac{d\gamma B(t)}{dt} + \cdots + \frac{1}{n!} \int_0^T t^n \mathcal{G}(t) dt + \cdots \right] \]

From the given sum, it can be seen that the \( n \)th integral expression corresponds to the \( n \)th gradient moment with respect to time. The net phase shift observed at an echo in MRI can thus be made independent of the \( n \)th order motion if the \( n \)th gradient moment is zero at that time. Zeroth moment nulling corresponds to the common rephasing of static spins, employed in all standard MRI pulse sequences.

Gradient-design strategies aimed at nulling certain higher gradient moments are often referred to as gradient motion rephasing, gradient moment nulling or, simply, flow compensation. Gradient schemes for first-order motion (velocity) compensation are fairly simple in their design (Figure G.36), but the nulling of higher-order gradient moments tends to become increasingly complicated and time consuming to execute.

Figure G.36a illustrates a general example of a three-lobe gradient waveform that will result in nulling of the zeroth as well as the first gradient moment, i.e. static spins and spins with constant velocity will be rephased. However, a number of alternative gradient schemes can be designed to accomplish the same result. Figure G.36b shows examples of slice-selection and frequency-encoding (readout) gradient waveforms for first-order motion compensation in a normal gradient-echo-type pulse sequence. The net phase shift observed at the centre of the readout period (arrow) is independent of zeroth and first order motion components, i.e. static spins and spins moving at a constant velocity will be rephased. Any phase shifts resulting from higher-order motion components along the slice-selection or frequency-encoding directions will, however, remain.

Related Articles: Flow compensation, Pulse sequence, Gradient field

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**Gradient spoiling**

(Magnetic Resonance) Gradient spoiling is used to destroy the coherence of an MR-signal by adding a strong uncompensated gradient in a pulse sequence after signal sampling. As a consequence of this spoiler gradient the spins at different locations will experience a variety of magnetic field strengths and thus will precess at different frequencies. Therefore the spins will quickly become dephased and the net magnetisation in the transverse direction will be reduced.

Gradient spoiling is not a very efficient way to destroy the transverse magnetisation. In particular, spins close to the isocentre of the magnet will be less affected by gradient spoiling than spins distant to the isocentre.

**Related Article:** Gradient field

**Grand-daughter radionucleus**

(Nuclear Medicine) In a decay chain the initial radioisotope disintegrates into another radioactive isotope. The second radioisotope is called the daughter radioisotope or decay product. The daughter radioisotope will eventually decay to a second decay product, or 'grand-daughter' radioisotope. This chain will continue until one of the decay products is stable. In nuclear physics, a specific part of a decay chain can be of clinical and/or research interest (the parent radioisotope can be chosen subjectively). For instance \(^{99}\text{Mo}\) disintegrates to the meta-stable state of \(^{99m}\text{Tc}\). \(^{99m}\text{Tc}\) decays into the grand-daughter radionuclide \(^{99}\text{Tc}\). \(^{99}\text{Tc}\) is also radioactive but emits no \(\gamma\) rays suitable for imaging, neither does \(^{99}\text{Mo}\). It is therefore important to elute carrier-free \(^{99m}\text{Tc}\) to avoid unnecessary radiation dose contribution to the patient.
The relationship between parent and grand-daughter activation is described by the Bateman equations.

**Related Articles:** Parent radionucleus, Daughter radionucleus, Bateman equations

**Graphite**

*(General)*

<table>
<thead>
<tr>
<th>Property</th>
<th>Value</th>
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</thead>
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<td>Melting point</td>
<td>3948 K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>4300 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>2090–2230 kg/m³</td>
</tr>
<tr>
<td>Specific heat capacity</td>
<td>0.71 kJ/kg/K</td>
</tr>
</tbody>
</table>

Graphite is an allotrope of carbon; it consists of layers of carbon atoms arranged in a hexagonal lattice. It is a semimetal that conducts electricity. Graphite is very commonly used in pencils, under the name ‘lead’. It is found naturally in three forms; flake, lump and amorphous, and it can also be produced synthetically.

**Medical Applications:** Graphite can be used as an alternative material to water in calorimeters used to measure the absorbed dose from radiotherapy linear accelerators.

**Related Articles:** Carbon, Calorimeter, Dosimeter, Lattice, Radiotherapy, Water calorimeter

**GRASE (gradient and spin echo)**

*(Magnetic Resonance)* See Gradient and spin echo (GRASE)

**Grating lobes**

*(Ultrasound)* Grating lobes is an unwanted effect that arises as a consequence of spatial sampling of the sound beams when a transducer array is used. Normally an array transducer should have only a main lobe extending perpendicular from the transducer face.

**Gray**

*(Radiation Protection)* This is the SI unit for absorbed dose and kerma of ionising radiation, and is abbreviated to Gy:

\[ 1 \text{ Gy} = 1 \text{ J/kg} \]

The Gray was defined in 1975 in honor of Louis Harold Gray (1905–1965) an English physicist whose work concentrated mostly on the effects of radiation on biological systems.

**Related Articles:** Kerma, Calculation of absorbed dose

**Green’s function**

*(Ultrasound)* A Green’s function is a solution to the inhomogeneous differential equation:

\[ \nabla^2 g - \frac{1}{c^2} \frac{\partial^2 g}{\partial t^2} = -\delta(r - r_0) \]

where the source term on the right hand side describes a point source located at \( r_0 \). If the medium is unbounded and the source simple harmonic, then the field at \( r \) is \( g(r, r_0) = e^{i\omega k(4\pi R)} \), with \( R = |r - r_0| \), \( k = \omega/c \), \( \omega = 2\pi f \), and \( f \) the frequency. At a given instant, this function describes an oscillation with a period in space that is related to the ratio of the frequency of the source and the propagation velocity, extending outward from the source point with decreasing amplitude as \( 1/R \). Green’s functions are used in calculations of sound fields and are advantageous to use as any source function can be described as a summation of point sources.

**Grenz rays**

*(Radiotherapy)* The term Grenz-ray therapy is used to describe treatment with beams of very low-energy x-rays produced at potentials below 12–20 kVp. Their half-value thickness lies below 0.04 mm Al. A negligible part of the Grenz rays are absorbed in air, i.e. their half-value layer at 4 kVp is approximately 11 cm. When using a different focus-skin distance therefore the inverse square law is not so readily applicable. Another disadvantage of Grenz rays, in particular the very soft ones, is that they are so strongly absorbed that it is difficult to measure them and to control the dose accurately. Because of the very low depth of penetration such radiation is no longer used in radiotherapy.

**Grey levels**

*(General)* A digital image is displayed using a grid of picture elements or pixels. In nuclear medicine each pixel contains the total number of counts at a particular x–y location and is displayed at a brightness level appropriate to this value. In black-and-white or greyscale displays the brightness is represented by varying degrees of white over a number of grey levels.

An image display system is characterised by the number of grey levels it can display. An 8-bit greyscale, for example can potentially display 256 (2^8) different grey levels. This is limited however by the capability of the human eye to distinguish between the levels and the limits imposed by image noise.

**Related Article:** Image display

**Greyscale**

*(Ultrasound)* Greyscale describes the range of greys used for conventional ultrasound images where, by convention, higher intensity echoes are shown as brighter greys. The 2-D image is known as the B-mode (B for brightness) or the grey (US grey) scale image (Figure G.37). Very weak echoes are shown as black and high intensity echoes as white. The distribution of grey levels in the image should be optimised to show good contrast resolution of the tissue under investigation. The appearance of the grey levels in the image depends on a number of instrument settings including power, gain, dynamic range, post-processing and harmonic imaging.

![FIGURE G.37 B-mode image of a kidney. The scale shown on the R of the image shows the gradation of the greyscale.](image-url)
Grid, Bucky

*Diagnostic Radiology* Grid is the common name of the device that is used in x-ray imaging to selectively attenuate the scattered radiation between the patient’s body and the image receptor as illustrated on Figure G.38 (the device is also known as anti-scatter grid).

The grid includes thin lead strips (absorbent lamellas) separated with non-absorbent plastic (or carbon fibre) material. The orientation of the lead strips allows for the primary x-ray beam to pass through with minimal absorption, while the majority of the scattered x-rays are absorbed, as the direction of their pathways differs significantly from the primary x-ray beam. This way mainly primary x-rays reach the detector (film).

Usually the lead strips have linear or crossed construction. The more often used linear grids have one set of parallel strips (parallel grids and focused grids). Most of the articles and diagrams here refer to the linear grids. Other grids use two set of crossed strips (crossed grids and rhomboid grids).

Bucky diaphragm is the classic name for what is more commonly known as a grid used to selectively absorb scattered radiation in x-ray imaging. It is named for Dr. Gustave Bucky of Germany who developed the first grid or ‘diaphragm’ in 1913.

A major advancement was made by Dr. Hollis Potter in 1920 who developed a method for moving the grid during the exposure to blur out the undesirable images of the grid lines.

This was known as the Potter–Bucky diaphragm. Different designs of such grids had been later introduced, as for example in the Lisholm scull stand, etc.

The use of the names has changed over the years. Now the ‘Bucky diaphragm’ is most commonly known as a grid and the ‘Potter–Bucky diaphragm’ is known as a ‘Bucky’ or a Bucky mechanism.

**Related Articles:** Focused grid, Grids crossed, Grid efficiency

### Grid efficiency

*Diagnostic Radiology* Grid efficiency is the combination of two factors. One is the ability to attenuate scattered radiation, which is

![Grid function](Graphs courtesy of Sprawls Foundation, www.sprawls.org)

**FIGURE G.38** Grid reducing scattered radiation concept. (Graphs courtesy of Sprawls Foundation, www.sprawls.org)

its purpose, and the other is the undesirable attenuation of the primary radiation. Both of these are related to several factors but the predominant factor affecting grid efficiency is grid ratio as illustrated on Figure G.39.

Grid ratio (usually with values from 5:1 to 16:1) is a ratio between height (thickness) of the strips and width of the interspace. In general, the efficiency of a grid increases with grid ratio because the attenuation of the scattered radiation increases more than the attenuation of the primary radiation.

**Related Articles:** Bucky diaphragm, Grid

### Grid ratio

*Diagnostic Radiology* The grid ratio of an anti-scatter grid is a design characteristic of a grid as illustrated in Figure G.40. Higher ratio grids have increased attenuation of scattered radiation and produce improved image contrast at the cost of a reduced sensitivity.

![Grid ratio concept and formula](Graphs courtesy of Sprawls Foundation, www.sprawls.org)

**FIGURE G.40** Grid ratio concept and formula. Some typical values could be $t = 2 \text{ mm}, d = 0.25 \text{ mm}$, producing grid ratio 8:1 (the thickness of the lead strips is on the order of 0.05 mm). (Graphs courtesy of Sprawls Foundation, www.sprawls.org)
The grid ratio is given in values as 5:1, 12:1, etc. These values are usually given in grid specification together with the density of the lead strips (or lamellas) – for example 40, 28L/mm, etc. This way one grid specified as Pb 12/40 means that it has lead strips with density of 40L/mm and grid ratio 12:1.

**Related Article:** Grid efficiency


**Grid-control**

(Diagnostic Radiology) See Grid-controlled x-ray tube

**Grid-controlled x-ray tube**

(Diagnostic Radiology) The grid-controlled x-ray tube is used for creation of sequences of very short x-ray pulses (as for cine-radiography). This is achieved by applying small negative voltage to the Wehnelt electrode (the cathode cup). Increased negative charge of the Wehnelt electrode creates strong space-charge effect, what blocks the thermal electrons and stops the anode current (in some tubes the Wehnelt electrode is maintained charged at about ~2kV). Applying control pulses to the Wehnelt electrode leads to very effective control of the beam of thermal electrons – hence producing sequences of 1 ms x-ray exposures. These grid-controlled tubes are rarely used at present, as the new medium-frequency high voltage generators control very effectively the short exposures.

The name of this tube is probably related (by analogy) with the electron vacuum tubes (triodes, etc.), where a special electrode (a metal grid) is used to control the anode current.

**Related Articles:** Cathode, Space-charge effect, Filament current, Wehnelt electrode

**Grids, crossed**

(Diagnostic Radiology) In comparison to the linear anti-scatter grids, the crossed grids (also known as crisscross) have two set of crossed lead strips. These can be perpendicular or at various angles one to the other (e.g. rhomboid grid). Crossed grids are more expensive, but at the same time very efficient at removing scatter radiation. For example, a crossed grid with grid ratio 5:1 removes scatter similarly to a linear grid with ratio 12:1 (while using similar radiation dose).

**Related Article:** Grid, Bucky


**Gross tumour volume (GTV)**

(Radiotherapy) The gross tumour volume (GTV) describes the full extent of visible malignant growth to be treated with radiotherapy. It can be determined using many different imaging techniques such as CT and MRI. As the different imaging techniques vary in their specificity, often many imaging techniques will be used together with image fusion/registration techniques. Margins will be added to the GTV to form the clinical target volume (CTV) and the planning target volume (PTV). In some cases, such as post-operative radiotherapy, there may be no GTV defined (Figure G.41).

The use of GTV as a planning volume was proposed by the ICRU in Report 50 (with addendum 62). This report provides a common framework on prescribing, recording and reporting therapies, with the aim to improve the consistency and inter-site comparability. It details the minimum set of data required to be able to adequately assess treatments without having to return to the original centre for extra information.
<table>
<thead>
<tr>
<th>Element</th>
<th>Protons in the Nucleus</th>
<th>Neutrons in the Nucleus</th>
<th>Nuclear Spin I</th>
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<th>$\omega$ at 1.5T (MHz)</th>
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**H and D curve**

*(Diagnostic Radiology)* The $H$ and $D$ curve, also known as the characteristic curve for radiography film, is the graph representing the relationship between optical density and exposure. It is named for the Swiss chemist Ferdinand Hurter (1844–1898) and English chemist Vero C. Driffield (1848–1915) who developed this concept for describing the characteristics of photographic emulsions.

A specific feature of the $H$ and $D$ or characteristic curve is the logarithmic scale used for exposure. The slope of the curve at each point represents the contrast transfer characteristics of the film. For more detail see the article on *Characteristic curve*.

**Related Articles:** Characteristic curve

**Hadron therapy**

*(Radiotherapy)* Hadron therapy involves the use of proton, neutron and ion beams (such as carbon) for radiotherapy treatment. Charged hadron beams have a depth dose dependence characterised by the Bragg peak due to the dominance of collision energy losses described by the Bethe–Bloch equation. This means that, for a certain energy, the dose distribution has a maximum at a certain depth. Often, the energy of the beam is modulated to smear this depth out producing a spread out Bragg peak (SOBP). Neutrons (being neutral in charge) have a depth dose characteristic similar to x-ray beams. There is evidence that hadron beams may have higher radiobiological effect (RBE) than x-ray or electron beams.

**Abbreviations:** RBE = Relative biological effect and SOBP = Spread out Bragg peak.

**Related Articles:** Charged particle therapy, Heavy particle beams, Ion therapy, Proton therapy

**Haemodynamic response function**

*(Magnetic Resonance)* In functional magnetic resonance imaging the temporal variation of the BOLD signal associated with increased neuronal activity is known as the haemodynamic response function. A typical haemodynamic response function for a brief stimulus is shown in Figure H.1. The main features are as follows:

- **Dip:** some studies have reported an initial dip, whilst others have failed to replicate the finding. This remains a controversial issue.
- **Positive gradient:** the MR signal rises due to an increase in the oxyhaemoglobin to deoxyhaemoglobin ratio.
- **Peak bold effect:** it typically occurs after around 5 s. If the stimulus was of long duration (as in many simple block designs) a plateau would be expected instead of a peak.
- **Negative gradient:** after the cessation of the stimulus, the oxyhaemoglobin to deoxyhaemoglobin ratio begins to relax back towards its baseline level.
- **Undershoot:** an effect believed to stem from the fact that increased venous blood volume normalises at a slower rate than increased venous blood flow, causing a relative increase in the deoxyhaemoglobin level.

**Related Articles:** fMRI (functional magnetic resonance imaging), Oxyhaemoglobin, BOLD, Block design


**Half acquisition single-shot turbo spin echo (HASTE)**

*(Magnetic Resonance)* HASTE is a special form of fast spin echo sequence, which combines single shot imaging with partial Fourier encoding. The latter exploits the symmetry of the raw data in $k$-space requiring the acquisition of part of the $k$-space data. This increases the temporal resolution of the pulse sequence.

**Related Articles:** Fast spin echo (FSE), Half Fourier imaging (HFI), Rapid acquisition relaxation enhancement (RARE)

**Half Fourier imaging (HFI)**

*(Magnetic Resonance)* One common goal in magnetic resonance imaging (MRI) is to reduce the image acquisition time. Conventional MRI images are generated by a two-dimensional Fourier-transform of the $k$-space data. Under the assumption that the image is a real function, the sampled $k$-space contains redundant information.

Consider the Fourier transform connecting the image space and the $k$-space, given by

$$ S(k) = \int r(x) \exp(-i2\pi k \cdot x)dx $$

(H.1)

where

- $S(k)$ is the signal at the $k$-space position $k = (k_x, k_y)$
- $r(x)$ is the image intensity at the image position $x = (x, y)$

If $r(x)$ is real, then

$$ S(k) = S^*(-k) \Leftrightarrow \int r(x) \exp(-i2\pi k \cdot x)dx = \left( \int r(x) \exp(i2\pi k \cdot x)dx \right)^* $$

(H.2)

since $r^*(x) = r(x)$, where $r^*(x)$ is the complex conjugate of $r(x)$.

In other terms, the $k$-space contains redundant information and a full-resolution image could be reconstructed by only sampling half of $k$-space and reconstructing the other half by Equation H.2. Omitting to sample parts of the $k$-space as described earlier is a technique called half-Fourier imaging or partial-Fourier imaging.
However, if the image is complex (e.g. due to the influence of $B_0$ inhomogeneities), Equation H.2 does not hold and reconstructing artefact-free images from partial-Fourier imaging requires more advanced reconstruction schemes. Various methods for doing the reconstruction have been suggested, for homodyne and projection onto convex subsets reconstruction. For details regarding these methods, see references.

**Related Article:** $k$-space


**Half Fourier turbo spin echo**

(Adapted from *Magnetic Resonance Imaging*) See *Half acquisition single-shot turbo spin echo* (HASTE)

**Half value layer (HVL)**

(Adapted from *Diagnostic Radiology*) Half value layer (HVL) is the most frequently used quantity or factor for describing both the penetrating ability of specific radiations and the penetration through specific objects. HVL is the thickness of material penetrated by one half of the radiation and is expressed in units of distance (mm or cm) referred to that specific material.

Increasing the penetrating ability of a radiation increases its HVL. HVL is related to, but not the same as, average photon range. There is a difference between the two because of the exponential characteristic of x-ray attenuation and penetration. The specific relationship is

\[
\text{HVL} = 0.693 \times \text{Average range} = 0.693/\mu.
\]

This shows that the HVL is inversely proportional to the attenuation coefficient. The number, 0.693, is the exponent value that gives a penetration of 0.5.

HVL can be used as an indicator for the total filtration of an x-ray tube (Figure H.2). Special tables and graphs (x-ray specific) are used to transfer the HVL value into total filtration (mm Al equivalent).

![Half Fourier turbo spin echo](image)

**FIGURE H.2** A typical curve of HVL measurement of an x-ray tube. The 50% line shows an HVL of 2.65 mm Al equivalent.

**Half wave rectification**

(Adapted from *Diagnostic Radiology*) See Rectifier

**Half-life of radionuclides**

(Adapted from *Nuclear Medicine*) The half-life of a radionuclide is the time taken to reduce its activity to half of its initial value. The half-life is unique for a given radionuclide and is usually denoted by $T_{1/2}$.

If initially there are $N_0$ nuclei in a radionuclide, then at time $t = T_{1/2}$ there is 50% of $N_0$ nuclei remaining:

\[
N_t = \frac{N_0}{2}
\]

$T_{1/2}$ is related to the decay constant $\lambda$ by

\[
\lambda = \frac{0.693}{T_{1/2}}
\]

The half-life of a radionuclide can therefore be used to calculate its activity at any given time, provided the initial activity of the radionuclide is known.


**Half-life of radionuclides in medicine**

(Adapted from *Nuclear Medicine*) The half-life of a radionuclide is the time taken for the activity to decay to half of its original value. The ideal half-life for a radionuclide depends on the purpose, i.e. if the radionuclide is used for imaging or radiotherapy. Radionuclides with a short half-life are preferable when imaging, since most of the activity will decay during image acquisition and no unnecessary absorbed dose is received by the patient. The principal limitations for using short-lived radionuclides are production and quality assurance related. In examinations using short-lived radionuclides the patient must receive a live feed of newly produced radionuclides from an accelerator or cyclotron, otherwise the activity will decay while transporting from the production site to the examination room. The live feed does not allow any complicated chemical labelling or extensive quality control of the radiopharmaceuticals. Another factor to consider is the redistribution rate, i.e. how long until the radionuclide enters the biological process of interest and for how long the process goes on. The radionuclide used for clinical imaging is therefore an optimisation between patient dose, the biological redistribution rate, production and quality assurance limitations.

For radionuclides used for therapy purposes there are a number of extra variables to take into consideration. For example, the redistribution time for large proteins is generally in the order of 1–2 days and therefore these need to be labelled with a radionuclide with corresponding half-life. However, as for imaging the technical aspects prevent the use of any radionuclide with half-life less than ~1 h.

**Hardening agent**

(Adapted from *Diagnostic Radiology*) A hardening agent is a chemical component of the fixer solution used in radiographic film processing to shrink and harden the emulsion layer after the processing actions are complete. Aluminium chloride is typically used as a hardener.

**Hardening, beam**

(Adapted from *Diagnostic Radiology*) See Beam hardening
**Harmonic imaging**

(Ultrasound) Harmonic imaging is an imaging modality that capitalises on non-linear effects to produce images with improved contrast and/or resolution. The first clinical application of harmonic imaging was to improve the detection of contrast agents. The approach was simply to transmit a pulse at one (centre) frequency, but have the detection at twice that frequency using a bandpass filter. This assumes a transducer that is wideband enough, and also that the pulses need to be fairly narrowband in order to avoid spectral leakage from the fundamental to the harmonic.

A bi-effect was also that the system was sensitive enough to pick up the harmonics generated by non-linear distortion. The idea had been suggested earlier, but had not gained clinical practice. Harmonics are generated at high acoustic pressures (see non-linear propagation), which naturally is in the centre of the sound beam. Thus the beam of the detected sound is narrower than the transmitted, and consequently the lateral resolution is improved. An advantage in cardiac investigations is also that multiple reflections between the transducer face and the ribs, for instance, are not detected in harmonic imaging mode. In normal (fundamental) imaging these multiple reflections turn up as a ‘haze’ in the image. However, the acoustic pressure does not build up here to the levels where harmonics are generated, and this unwanted noise is suppressed in harmonic mode. In other words, the contrast resolution has improved.

**HASTE (half acquisition single-shot turbo spin echo)**

(Magnetic Resonance) See Half acquisition single-shot turbo spin echo (HASTE)

**Hazard**

(Radiation Protection) Hazard, or health hazard, is defined as anything that has the capacity to cause harm to a human being. Such harm can be through physical injury, biological damage or other health detriment (adverse effect) in the form of cancer or other disease. Examples of hazards include fire, chemicals, electricity (electrocution), water (drowning), and ionising and non-ionising radiation exposure. If the hazard relates to radiation exposure, it may be called a radiation hazard.

**Related Articles:** Adverse effects, Risk assessment

**HC (homogeneity coefficient)**

(Radiotherapy) See Homogeneity coefficient (HC)

**HDR**

(Radiotherapy) See High dose rate

**Head, pitch and roll**

(Radiotherapy) There are three main axes about which movement can happen – vertical, longitudinal and lateral. The movement can be either a displacement or a rotation about the axis. When the motion is a rotation, the terms head, pitch and roll are used for the three axes as illustrated in Figure H.3. Occasionally the term ‘head’ is replaced by ‘yaw’. These terms can be used in the radiotherapy context when checking the gantry set-up of a linear or for the rotational set-up error of a patient.

**Related Article:** Set-up error

**Head coil**

(Magnetic Resonance) A head coil is an RF coil used in brain and head imaging.

Head coils are volume coil designs, providing good signal uniformity throughout the imaged volume. Head coils can be designed in a transmit/receive (T/R) configuration. Use of a T/R coil design avoids the need for RF excitation by the body coil and reduced SAR (specific absorption rate).

**FIGURE H.3** An illustration of the three axes and the relevant term relating to rotation about that axis.

**FIGURE H.4** Head coil.

The most commonly used head coils (Figure H.4) are circularly polarised head coils such as a birdcage coil or multiple array head coils such as an eight channel coil.

**Health hazard**

(Radiation Protection) Health hazard, or hazard, is defined as anything that has the capacity to cause harm to a human being. Such harm can be through physical injury, biological damage or other health detriment (adverse effect) in the form of cancer or other disease. Examples of hazards include fire, chemicals, electricity (electrocution), water (drowning) and ionising and non-ionising radiation exposure. If the hazard relates to radiation exposure, it may be called a radiation hazard.

**Related Articles:** Adverse effect, Risk assessment

**Health Protection Agency (HPA)**

(Radiation Protection) In 2005 the National Radiological Protection Board merged with the Health Protection Agency (HPA)
to become an independent body that assesses the impact of all sources of radiation on public health in the United Kingdom, and advises on the safe use of radiation.

The HPA also monitors communicable diseases, performs epidemiological studies and advises on infection and chemical hazards. The HPA has three core centres: Centre for Emergency Preparedness and Response, Centre for Radiation, Chemical and Environmental Hazards and Centre for Infections.

The HPA runs various training programs for health professionals and carries out simulations to test public health systems in times of emergency.

Heat unit

(Diagnostic Radiology) Heat unit (HU) is a practical measure introduced in the past to measure the heat capacity of the x-ray tube anode. This measure continues to be used, but as it is directly related to energy, it is often replaced by Joules (J). By definition 1 HU = 1.4 Joules.

HU has been introduced when most high-voltage generators have been single phased two pulsed type. Due to this reason HU refers to the effective kV (i.e. the root mean square kV = 0.71 kVp) and the average mAs. Effective kV (or r.m.s. kV) is used to represent the constant kV with the same effect as the original pulsating kVp. This way the HU for such generators are calculated very easy:

\[ 1 \text{HU} = 1.4 \text{kVeff mA s} = 1.4 (0.71 \text{kVp}) \text{mA s} = \text{kVp mAs} \]

The introduction of other types of generators (as three phased, constant potential, etc.) changed the representation of the kVp into constant kV (as the wave form pulsations are different), what requires introduction of specific coefficients related to the pulses of the kVp (i.e. the type of rectification). This way:

a. HU for three phase six pulse generator (when kVp are very close to constant kV) will be

\[ 1 \text{HU} = 1.4 \text{kVeff mA s} = 1.4 (0.96 \text{kVp}) \text{mA s} = 1.35 \text{kVp mAs} \]

b. HU for constant potential generator (as most medium frequency generators) will be

\[ 1 \text{HU} = 1.4 \text{kVp mAs} \]

The heat capacity of the anode of an x-ray tube is very often presented in HU, thus a good x-ray tube for computed tomography will have several millions heat unit capacity (e.g. 5 MHU).

Related Article: Voltage waveform


Hyperlink: www.sprawls.org

Heating

(Ultrasound) As ultrasound passes through tissue, some of its energy is absorbed in tissue and is converted to heat. Ultrasound induced heating is dependent on

- Intensity
- Tissue properties, e.g. absorption
- Heat clearance, conduction and perfusion
- Proximity to the transducer

Heating is a potential adverse effect for diagnostic ultrasound. It is known from tissue and animal studies that elevated temperatures can cause biological effects and that these are dependent on temperature rise and length of time at raised temperatures. It is known from in vitro studies that ultrasound at high diagnostic levels can raise temperatures in the order of 1°C–2°C. Studies of outputs, the heating effect and cell and animal experiments have been used to define safe levels at which to operate. This is particularly important when scanning embryos and foetuses since a largely normal population is screened at a time when rapid development occurs. Non-ultrasound heating studies have demonstrated teratogenic effects at temperature rises of 4°C for 5 min. Effects at lower temperatures are uncertain; other factors may be important, for example if the mother has raised temperature herself. The World Federation of Ultrasound in Medicine and Biology (WFUMB) have stated that systems producing a temperature rise of no more that 1.5°C can be used clinically without reservation.

Manufacturers, regulatory bodies and user groups have developed guidelines and recommendations to ensure that heating caused by diagnostic ultrasound is low and that outputs sufficient for effective imaging are permitted. The current output safety indices (ODS) include a Thermal Index displayed on systems as an indication of the relative heating risk.

Ultrasound probes are themselves prone to heating. In advanced matrix arrays, the high density of elements can cause temperature rises that required cooling, either by conduction to the rear of the transducer case or cooling by circulating fluid.

For therapeutic ultrasound devices, for example those used in physiotherapy, low temperature heating may be the desired mechanism used. In high intensity focussed ultrasound (HIFU) heating is used as the mechanism to destroy pathogenic tissue. The power and intensity are much higher than for diagnostic applications and the design of HIFU systems is markedly different from diagnostic systems and other therapeutic applications.

Related Articles: Thermal index, Intensity, HIFU


Heating, radiofrequency

(Magnetic Resonance) See \textit{Radiofrequency heating}

Heating, resistive

(Diagnostic Radiology) See \textit{Filament}

Heavy charged particle stopping power

(Radiotherapy) The stopping power may be subdivided into collision and radiative stopping power. The latter depends on the inverse square of the mass of the particle for a given particle velocity and therefore it becomes practically insignificant for particles other than electrons and positrons. The stopping power for a heavy particle in any medium is given therefore by

\[ -\frac{dE}{dx} = \frac{4\pi\kappa z^2 e^2 n}{m c^2 \beta^2} \ln \frac{2mc^2 \beta^2}{I(1-\beta^2) - \beta^2} \]

where

- \( \kappa_0 = 8.99 \times 10^8 \text{N m}^2/\text{C}^2 \)
- \( z \) is the atomic number of the heavy particle
- \( e \) is the magnitude of the electron charge
- \( n \) is the number of electrons per unit volume in the medium
- \( m \) is the electron rest mass
- \( c \) is the speed of light in vacuum
- \( \beta = ve/c \) is the speed of the particle relative to \( c \)
- \( I \) is the mean excitation energy of the medium

Examination of the equation shows that stopping power is a function of the charge \( ze \) and velocity \( V \) of the heavy charged particle but
therapy equipment in several ways, including the following.

Heavy ions in any medium can be written as

\[ \frac{dE}{dx} = \frac{5.08 \times 10^{-31} \text{eV}}{\beta^2} \left[ F(\beta) - \ln I_0 \right] \text{MeV/cm} \]

where

\[ F(\beta) = \ln \frac{1.02 \times 10^6 \beta^2}{1 - \beta^2} - \beta^2 \]

**Related Articles:** Stopping power, Collision mass stopping power, Electron stopping power, Mass collision stopping power, Mass radiative stopping power, Mass stopping power, Restricted mass collision stopping power, Restricted stopping power

**Heavy ions**

*(General)* A heavy ion is one with a mass greater than an alpha particle. They can be used in charged particle therapy. Please see related articles for more information

**Related Articles:** Charged particle therapy, Hadron therapy, Ion therapy, Proton therapy, Heavy particle beams

**Heavy metal filter**

*(Radiotherapy)* Heavy metal filters are commonly used in radiotherapy equipment in several ways, including the following.

1. **Beam flattening filter** – The intensity of an x-ray beam generated in mega-voltage radiotherapy equipment such as linear accelerator has a Gaussian distribution across the radiation field. A conical shape metal flattening filter of several cm in thickness of copper or lead alloy is used to flatten the beam so as to achieve a uniform beam intensity across the field. The flattening filter normally sits on a turret, which rotates to fit different flattening filters in position when beam energy is changed, and is located between the x-ray target and the ionisation chamber. Modern mega-voltage radiotherapy equipment such as Tomotherapy and the latest generation of linear accelerators do not require the use of flattening filters. This is because uniform beam intensity is no longer required in advanced treatment techniques such as intensity modulated radiotherapy and intensity modulated arc radiotherapy.

2. **Wedge filter** – See *Wedge*

3. **Hardening filter** – Hardening filters are commonly used in x-ray therapy unit operating in the kV range, such as superficial x-ray therapy (see *Superficial therapy*). The main purpose is to increase the penetration power of the x-ray beam to meet the required treatment depth. The filter cuts down the amount of low energy x-ray component and this effectively increases the mean energy of the beam, but at the expenses of lower beam intensity.

4. **Electron beam flattening filter** – This is a thin rather than heavy metal foil made of gold or copper used for flattening radiotherapy electron beams. This type of filter is also known as electron beam scatterer. Electron flattening filter also sits on the same turret as the x-ray flattening filter for automatic selection to meet treatment requirement. The main disadvantage of using electron scatter is that it generates bremsstrahlung radiation which is undesirable in electron beam treatment. The amount of bremsstrahlung radiation produced is proportional to electron energy. In order to reduce x-ray contamination, a double scatter design is used in high energy electron beam treatment.

**Heavy particle beams**

*(Radiotherapy)* Heavy particle beams are beams of neutrons, protons, light ions (such as carbon) used in hadron therapy. Charged heavy particle beams are generated using particle accelerators. These accelerators are substantially larger than the more familiar electron linac due to the higher energies needed to penetrate tissues to the required depths and the high magnetic rigidity of the much heavy hadron particles. Neutron beams are generated using three methods: hadron accelerators with a light target to generate neutrons; radioactive sources; and nuclear reactors.

**Related Articles:** Charged particle therapy, Hadron therapy, Ion therapy, Proton therapy

**Helical artefact**

*(Diagnostic Radiology)* Helical (spiral) CT suffers in general from the same artefacts as sequential (axial) CT. However, there are effects, or artefacts, which are seen only helical CT, and these are due to the spiral interpolation process which gives rise to data inconsistencies at different angular orientations. These artefacts
Helical pitch (CT)

- **Diagram (Figure H.8)**: Pitch in CT scanning (from left to right: pitch 1, pitch 2, pitch 0.5).

**Helical pitch (CT)**

(Diagnostic Radiology) In helical scanning the speed that the patient support table moves through the gantry during the scan can be varied. The ratio of the table feed per x-ray tube rotation to the x-ray beam length in the z-axis is referred to as the helical pitch, pitch ratio, or simply, pitch. If the table feed per gantry rotation is equal to the length of the x-ray beam, the pitch is equal to one (Equation H.3). This is often referred to as contiguous scanning:

\[
\text{Helical pitch} = \frac{\text{(Table feed per rotation)}}{\text{(z-axis x-ray beam length)}} \quad (H.3)
\]

For faster or slower couch speeds, i.e. non-contiguous scanning, the pitch value increases and decreases respectively (Figure H.8). The advantage of using a high pitch is a reduction in exam time, although on single slice scanners it also results in a broader slice thickness or slice sensitivity profile. Another advantage of high pitch on single slice scanners is dose reduction.

Higher pitches result in an increase in reconstruction artefacts, and so pitch values greater than 1.5 are seldom used. Because of the high gantry rotation speeds and increased detector array lengths on MSCT scanners, use of high pitches is rarely necessary as scan times are sufficiently short at pitch values of around one or less.

**Related Articles:** Helical scanning, Slice thickness, Image artefact

**Helical scanning**

(Diagnostic Radiology) In computed tomography (CT), the terms helical and spiral scanning are synonymous. Helical scanning was introduced into clinical practice in 1990. It was enabled by the implementation of slip-ring technology on CT scanners, which removed the need for power and signal cables between the stationary and rotating parts of the gantry, and so allowed continuous rotation of the x-ray tube and detectors in one direction. When this is coupled with simultaneous couch movement through the gantry during the scan, a volume of attenuation data is acquired. The introduction of helical scanning revolutionised CT, as it reduced the duration of a scan, thereby decreasing artefacts associated with voluntary and involuntary patient motion.

The filtered back projection process, commonly used to reconstruct images in CT, requires attenuation data for multiple angular positions in the scan plane (Figure H.9). When scanning in helical mode the angular data set is non-planar as the x-ray beam describes a helical path around the patient (Figure H.10). It is therefore necessary to interpolate the acquired data in order to obtain a planar data set for image reconstruction.

Different types of interpolation algorithms are used and these give rise to different slice thicknesses (slice sensitivity or z-sensitivity profiles). The slice sensitivity profile is widened to a greater or lesser extent depending on the type of algorithm used. The slice sensitivity profile is also affected by the helical pitch, particularly on single slice scanners, where higher pitches lead to wider profiles.

Besides its speed, helical scanning has the advantage that slices can be reconstructed at any arbitrary position along the scan axis (z-axis). It is common to reconstruct overlapping images, i.e. where the reconstruction interval is less than the slice width, and this

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**Image 94x335 to 274x515**

Single slice helical CT scan of a cone: images at different z-axis positions scanned with pitch = 2. Note this image is from a scanner ~1999. With current reconstruction techniques helical artefacts are reduced. (Courtesy Philips Healthcare, Best, the Netherlands.)

**Image 64x574 to 304x722**

Multislice helical artefact (windmill artefact) of a Teflon rod at 60° to the horizontal. (Graphs courtesy of ImPACT, UK, www.impactsan.org)

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**Table:**

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</table>

Helium

History: Helium was discovered in 1868 by Pierre Janssen and Norman Locker, who independently observed its spectral line in light from a solar eclipse.

Isotopes of Helium: Helium occurs naturally as two stable isotopes. By far more abundant is $^4$He, which makes up 99.99986% of all natural helium. The remaining 0.00014% consists of stable $^3$He, which is a product of tritium beta decay. Six other isotopes exist, all of which are unstable and have half-lives of less than 1 s.

Medical Applications: Hyperpolarised Gas Ventilation Imaging – $^3$He is used to view lung ventilation. The $^3$He gas is polarised using laser light before inhalation, and its distribution in the lungs can then be imaged using low flip angle or steady state magnetic resonance imaging. $^4$He is expensive due to its low natural abundance.

Guiding Lasers: A combination of helium and neon gases is a commonly used gain medium for guiding lasers, which produce light in the visible region of the electromagnetic spectrum. These are used for visual alignment of other lasers (e.g. excimer, dye, etc.) in medical procedures. Low-power helium-neon lasers are also used therapeutically in bio-stimulation and phototherapy.

Superconducting Magnet Coolant: Liquid $^4$He is used to cool the superconducting magnets, such as those used in MRI scanners, to temperatures below that at which the magnet metal becomes superconducting.

Neutron Detection: $^3$He gas is used in gas-filled ionisation chambers such as the Geiger–Müller tube, to enable detection of neutrons. An incoming neutron reacts with the gas to form tritium and protons, which can be detected and converted to an electrical signal.

Related Articles: Magnetic resonance Imaging, MRI, Cryogen, Geiger–Müller (GM) counters

Helium, liquid
(General) See Helium

Helmholtz coil
(Magnetic Resonance) A Helmholtz coil configuration is a pair of coils used to create uniform magnetic field in the centre of the space between them. The diagram shows two circular coils with radius $a$, and separated by a distance $d$ with each carrying equal currents $I$. The resulting magnetic field is a summation of the magnetic fields generated by each coil. It can be shown that in the case where the distance $d$ equals the radii $a$, the magnetic field is maximally uniform along the axis joining the coil centres. This configuration is called a Helmholtz coil (Figure H.11). The Helmholtz coil is useful in Magnetic Resonance imaging due its ability to create a uniform magnetic field.

Herring bone artefact
(Magnetic Resonance) The herring bone or ‘crisscross’ artefact is caused by spike noise in the raw data, typically due to RF interference. When Fourier transformed, the spike produces a regular pattern in the 2D frequency domain. This artefact can be reduced by using spectral shaping filters or by averaging multiple data sets.
series of high and low intensity stripes in one or more directions across the image. This leads to the herring bone or crisscross effect.

**Hertz (Hz)**

(General) Hertz (Hz) is the base unit of frequency in the International System of Units (Système international d’unités – SI). 1 Hz is 1 cycle per 1 s. The unit can be applied to various periodic events and is named after the German physicist Heinrich Hertz.

**Related Article:** Frequency

**Heterogeneity**

(Radiotherapy) A heterogeneity is a region in an irradiated medium which has a different composition (atomic number or density) to that of the surrounding medium. The human body is such a heterogeneous medium, being composed of a variety of tissues and cavities with different radiological properties than that of water.

The presence of these heterogeneities, with attenuation and scattering properties different to that of water, can affect the dose distribution by altering both the primary and scatter contributions to a point relative to that in a homogeneous water medium.

A non-water equivalent component in the human body can be referred to as a heterogeneity, an inhomogeneity, a tissue heterogeneity or a tissue inhomogeneity.

**Example:** Bone and lung are two examples of heterogeneities in the human body.

**Related Articles:** Tissue heterogeneity, Inhomogeneity correction factor

**Further Reading:** AAPM. 2004. Tissue inhomogeneity corrections for megavoltage photon beams. Report number 85, Medical Physics Publishing, Madison, WI.

**High activity sealed source**

(Radiation Protection) High activity sealed source (HASS) describes any sealed radioactive source that is covered by regulation introduced under European Directive 2003/122/EURATOM to provide additional safety and security in the management of such sources. From 1 January 2006 European member states were required to register any single source exceeding the following limits as HASS:

Radioisotope HASS threshold:

<table>
<thead>
<tr>
<th>Isotope</th>
<th>Activity Limit</th>
</tr>
</thead>
<tbody>
<tr>
<td>^55Fe</td>
<td>400 GBq</td>
</tr>
<tr>
<td>^60Co</td>
<td>4 GBq</td>
</tr>
<tr>
<td>^75Se</td>
<td>30 GBq</td>
</tr>
<tr>
<td>^85Kr</td>
<td>10 GBq</td>
</tr>
<tr>
<td>^90Sr</td>
<td>3 GBq</td>
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<tr>
<td>^103Cd</td>
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<td>^137Cs</td>
<td>20 GBq</td>
</tr>
<tr>
<td>^192Ir</td>
<td>10 GBq</td>
</tr>
<tr>
<td>^228Ra</td>
<td>2 GBq</td>
</tr>
<tr>
<td>^241Am</td>
<td>100 GBq</td>
</tr>
</tbody>
</table>

For example, in the United Kingdom, establishments are required to be authorised by the Environment Agency to hold radioactive substances. The requirements for holding medium/low activity sealed sources are covered under one licence that allows the organisation some flexibility to manage various sealed sources, with a broad set of activity limits and security requirements. However, a HASS must be individually registered with the agency, and requires an individual HASS licence which imposes additional security and management arrangements. The owner must provide a life-cycle management plan that details safe use, security and disposal of the source.

**High contrast**

(Diagnostic Radiology) There are several terms specifying the subject contrast (mainly used in x-ray radiography). The most often used ones are high contrast and low contrast.

High contrast (Figure H.12) is used to describe images with large differences between the optical densities of the adjacent objects. A number of synonyms are used in practice – as hard contrast or short scale contrast. The latter refers to the fact that x-ray films with small latitude present a limited (short) scale of gray levels.

Low contrast (Figure H.13) is used to describe images with small differences between the optical densities of the adjacent objects. A number of synonyms are used in practice – as soft contrast or
long scale contrast. The latter refers to the fact that x-ray films with big latitude present a large (long) scale of gray levels.

An objective way to describe the contrast (optical densities) in radiography is based on use of densitometer. The optical density (OD) values obtained by this device are logarithmic values of light transmission (OD = log10[1/I0]) through the film. Usually distinctive difference between the densities of two adjacent objects is considered to be 0.2 OD (this has been determined by many subjective assessments). Most often these 0.2 OD are achieved by difference of 58% between the x-ray exposures received by the two adjacent areas. Hence the maximal acceptable film fog is also 0.2 OD. Optical densities below 0.15 OD are difficult to be distinguished by a normal human vision.

In digital imaging, the window technique determines the type of contrast. A large window width (e.g. WW > 200) will present many gray levels and the image will be considered to be with low contrast (many anatomical structures will be seen, but with less differences between these) – Figure H.14. Correspondingly, small window width (e.g. WW < 100) will present few gray levels and the image will be considered to be with high contrast (only limited anatomical structures will be seen, but their contrast will be distinctive) – Figure H.15.

Terms used often in computed tomography (CT) practice are high contrast resolution (Figure H.16) and low contrast resolution (or detectability) (Figure H.17). The first one refers to the measure of resolution by test objects with high subject contrast between their inserts. This measure represents the spatial resolution of the system (Lp/mm). The second one refers to measures using test object with small subject contrast between their inserts. This measure is used to assess the noise and the contrast resolution of the system (limiting contrast, %).

**Related Articles:** Subject contrast, Low contrast, Film latitude, Contrast resolution, Window

**High counting rates**
(Nuclear Medicine) See Dead time

**High dose rate (HDR)**
(Radiotherapy, Brachytherapy)

*Dose Rates in Brachytherapy:* Different dose rates are used in brachytherapy, connected to different treatment techniques. ICRU, the International Commission on Radiation Units and Measurements, defined these dose rates in its Report No. 38 ‘Dose
and Volume Specification for Reporting Intracavitary Therapy in Gynaecology:

1. Low dose rate, LDR
   a. 0.4–2.0 Gy/h
   b. Traditional radium technique; 0.5 Gy/h, 60 Gy with treatment time 5 days
   c. Large amount of clinical data
   d. (NOTE! Ultra low dose rate 0.01–0.3 Gy/h!)

2. Medium dose rate, MDR
   a. 2–12 Gy/h
   b. More seldom used

3. High dose rate, HDR
   a. >12 Gy/h = 0.2 Gy/min
   b. Treatment times approx. 5–20 min (external beam therapy belongs here)
   c. Clinical data available

4. Pulsed dose rate, PDR
   a. Mimics LDR, using many small ‘HDR pulses’ during a longer treatment time
   b. Example: one pulse per hour during 24 h, 0.5 Gy per pulse given in 5 min; total dose 12 Gy per day.

The radiobiological effects in the tissues irradiated depend on the type of applicator used, on the fractionation scheme and on both dose and dose rate distributions. As stated in the ICRU Report 38: ‘the clinical experience accumulated with radium techniques cannot be applied to new irradiation conditions without careful consideration’. This includes consideration of both tumour effects and effects on normal tissues.

Abbreviation: ICRU = International Commission on Radiation Units and Measurements.

Related Articles: Brachytherapy, Dose rates in brachytherapy, see also articles under radiobiology


High energy electrons

(Radiotherapy) High energy electron beams in the energy range of 4–20 MeV are useful to treat superficial tumours and sometimes they are used in conjunction with photon beams either as a boost or a mixed-beam combination. The interaction of electrons with matter (collision) has specific features that allow the effective irradiation of relatively superficial tumours sparing the normal tissues beyond. In a treatment plan with photons the beam is exponentially absorbed by the tissues beyond the target depth while the electron beam, as constituted by charged particles, interacts with atoms and shows a finite range in matter limiting the dose to deep seated normal tissues. Disadvantages in the use of high energy electron beams are the rapid decrease of the beam uniformity moving away from the point of final collimation and the change in depth dose penetration when they traverse inhomogeneities.

High-frequency generator

(Diagnostic Radiology) This contemporary x-ray generator gradually replaces the classical high-voltage generator. The high frequency generator uses electrical current with 5–25 kHz frequency and due to this reason it is better named by some sources as medium frequency generator.

The principle of this new high-voltage generator is based on the known transformer equation:

\[
\frac{U}{f} = A n,
\]

where

- \(A\) is the cross section of the transformer core (mm²)
- \(n\) is the transformer ratio (based on the number of secondary/primary windings)
- \(f\) is the frequency of the secondary voltage \(U\)

For one x-ray high-voltage transformer \(n\) is constant (usually around 500). In this case it is obvious that increasing the frequency of the electricity \(f\) will allow for reducing the size of the transformer core. Classical iron-core transformers can not work with high frequencies (there will be too many losses due to their low magnetic permeability). However new ferrite-core transformers have much higher magnetic permeability and work with high frequencies. This allows great reduction of the transformer size. As a comparison, an iron-core high-voltage transformer (from classical x-ray generator) will weight around 300–600 kg (depending on its power), while a contemporary ferrite-core high-voltage transformer will be much lighter – approximately 25% of the volume and weight of the classical one.

The principal diagram of a medium (or high) frequency x-ray generator is given on Figure H.18. The mains power supply is rectified and than an DC/AC converter (also called inverter or chopper) is used convert the DC voltage to a series of pulses (usually with frequency of the order of 5–25 kHz, depending on the manufacturer). This high frequency electricity is transformed to high voltage by the special ferrite-core transformer, than is again rectified before supplying the x-ray tube with high voltage (also smoothed by special capacitor).

The DC/AC converter is normally thyristor-type, driven by external pulse generator. This combination of medium frequency and converter is further applied not only to the high-voltage transformer, but also to the filament circuit, anode rotation, etc. From this point of view a contemporary x-ray generator will have a number of ferrite transformers.

Although these XG are more expensive, they are very small, very efficient and produce extremely precise x-ray exposures. They do not have special requirements to the mains, do not need synchronisation, and produce \(U\), independent of \(I\). A very important feature of these high-voltage generators is that the high voltage can be controlled by controlling the frequency. This is obvious from the preceding equation, having on mind that for given x-ray generator, not only the transformer ratio \(n\), but also the size of the transformer \(A\) are constants. This way \(U\) is linearly proportional to \(f\). The converter can smoothly change the frequency, what leads to smooth change of the anode voltage (kV). Additionally these generators have much less pulses (i.e. less kVp ripple – see the article Voltage waveform).

Related Articles: High-voltage generator, High-voltage transformer, Capacitor-discharge generator, Pulse-less generator, Filament circuit, High-voltage circuit, Voltage waveform

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**FIGURE H.18** Block diagram of medium (high) frequency high-voltage x-ray generator.
High-intensity focused ultrasound (HIFU)
(Ultrasound) HIFU is a technique whereby pathogenic tissue is heated and destroyed by high energy focused ultrasound. In order to achieve accurate control of cell death, the transducer is designed to produce highly focused peak intensities, typically from 1000–2000 W/cm² in a precisely controlled volume. Transducer types include shaped single element transducers, a plane transducer with a lens and multi-element phased arrays. Typical frequencies are 0.5–5 MHz. These intensities produce high temperatures (typically 65°C) in a small contained volume, typically 1–2 cm length, 103 mm diameter. If larger volumes of tissue are to be ablated, then the transducer is swept over the larger volume. Monitoring of the heating is performed using MRI or ultrasound imaging.

Trials are under way to use the procedure in the treatment of cancers including those in prostate, liver, kidney breast and bone. The technique has been proposed to induce haemostasis in vascular trauma. One of the stated advantages of HIFU is that it has no cumulative adverse effect on surrounding tissue.


High kV technique
(Diagnostic Radiology) High kilovoltage x-ray imaging uses normally potentials higher than 100 kVp. It has special importance for imaging parts of the body with large absorption differences (e.g. bones and soft tissue). Due to this reason one of the most often uses of high-kV technique is in chest radiography. In this case the decreased image contrast associated with the high energy photons is of lesser importance as the contrast difference between ribs and lungs is very significant. Another area where high-kV x-ray imaging is useful is contrast examinations (due to the high absorption of the contrast media – iodine, barium, etc).

One of the advantages of this technique is reducing the patient absorbed dose (due to the increased penetration of higher energy photons). Another advantage is the reduced time of the exposure (very important also in the chest region). However in this energy region Compton scattering is the predominant interaction, what requires use of more effective anti scatter grids (high grid ratio). X-ray chest systems use focus to detector distance of the order of 200 cm and in this way there is smaller influence of the x-ray tube focal spot to the overall image resolution.

However, a number of contemporary measurements show that the new digital detectors produce better image quality (especially for chest images) with low-dose mid-range kV energy than high-kV imaging.

High-pass filter
(General) A high-pass filter is a filter that enhances high frequencies and reduces low frequencies. The high frequencies models high gradient regions in an image while low frequencies model larger structures. Therefore high-pass filters can be used to sharpen edges, e.g. to reveal structures hidden behind a uniformly distributed source.

High-voltage cable
(Diagnostic Radiology) The high-voltage cables are used to lead the high-voltage (HV) potential from the HV box (tank) to the x-ray tube. The HV cables have multi-wire copper leads with special rubber or plastic insulation (above 100 kV). A grounded metal shielding is placed above the insulation, and over it there is another protective cover (most often plastic) (Figure H.19). There are special requirements for the residual capacity of these heavy cables. At both ends the HV cables have special cable-ends (Figure H.20), connecting the HV to the x-ray tube. Silicon paste (or insulation oil) is used to assure the electrical safety of the connection. The HV cables have additional wires for cathode filament, anode, rotation, control circuitry, etc.

Related Articles: X-ray tube, High-voltage generator

High-voltage circuit
(Diagnostic Radiology) The two main circuits of the high-voltage generator (HVG) are the high-voltage circuit and the filament circuit. The high-voltage circuit of the HVG controls the high voltage (kV) supplied to the x-ray tube and thus controls the x-ray spectrum. In classical HVG this control is achieved primarily by changing the input voltage to the high-voltage transformer (HVT). A different method is used in high frequency generators (changing the frequency) – it will be described in the article on High-frequency generators.

The high-voltage circuit is diagrammatically presented as part of the high-voltage generator and high-frequency generator electrical circuits – see the eponymous articles.

The most primitive classical HVG consists of a high-voltage transformer (step-up transformer), which is connected to the single phase mains and supply the x-ray tube with un-rectified HV
In this case the x-ray tube (XT) serves also as a diode. This type of HVG is not used any more because the XT can easily be damaged by the negative wave of HV electricity. This is due to the fact that during this negative wave the already hot anode emits thermo-electrons, which bombard the thin cathode (which is now positively charged) and could easily destroy it.

Later the HV has been rectified with one power diode (single-wave rectification), known as single-pulse (or single phase) HVG. Further it has used a bridge-rectifying circuit with four diodes (full-wave rectification), known as two-pulse HVG. Although this HVG is more efficient it still produces 100% pulsations of the high voltage (the kV pulsations are explained in the article Voltage waveforms).

Such HVG is used for low power x-ray equipment (e.g. dental). To minimise the high-voltage requirements to the diodes (and the high-voltage cables insulation) the HV transformer is usually made of two halves (having a 0 potential at its central point) which supply, for example +60kV and −60kV to form 120kV potential difference between the anode and the cathode (Figure H.22). This method is widely used for all types of HVG and usually the 0 potential point (which is grounded) is the place where the tube current $I_a$ (mA) is measured.

The HVG using three-phase electrical power supply have much larger ‘tripled’ transformer and rectifying circuits (Figure H.23). Normally the secondary winding of the HVT is with star-connection. The three-phase bridge rectifier of this HVG (with six diodes) provides six pulses in one period, and the amplitude of pulsations is much smaller – approx. 14% pulsations. Due to this reason these HVG produce less soft radiation and are more efficient. The six-pulse HVG is used for power equipment (100–120kW). It can produce the same dose output as one two-pulse HVG, but for much shorter time.

Further there are three-phase HVG, in which HVT has two types of secondary winding connection – star and delta (Figure H.24). In this case the output pulsates are doubled (12-pulse HVG). This double frequency and 12 diodes bridge-rectifier produces kV close to constant potential (approx. 3.4% pulsations). This makes this HVG very effective, producing very short exposure times and suitable for high power output (up to 150kW).

Moreover special high voltage is applied to further reduce the pulsations of the HVG. This generator is known as pulse-less and will be described in the eponymous article.

Related Articles:
High-voltage generator, High-voltage transformer, High-frequency generator, Pulse-less generator, Voltage waveform


High-voltage control device
(Diagnostic Radiology) See Radiographic kV control

High-voltage generator
(Diagnostic Radiology) All electrical circuits which supply and control the electrical high-voltage energy to the x-ray tube are called high-voltage x-ray generator (sometimes also x-ray generator
The high-voltage transformer (HVT) is part of the high-voltage generator (HVG). The two main circuits of the high-voltage generator are the high voltage circuit and the filament circuit.

The first Roentgen's high-voltage generator has used single phase electrical supply and has included just one high-voltage transformer (HVT). Later high-voltage rectifiers and other additional circuitry have been added to the HVT. These have further developed to use three-phase electrical supply, control equipment, electronics, etc. All these design improvements have followed the basic principles of the first x-ray generator and have used the normal mains frequency of 50/60 Hz and HVT with iron core. Due to this reason these are known by a common name – conventional (or classical) high-voltage x-ray generators (HVG). Contemporary high-voltage generators use electrical frequency above 1 kHz and are known as high-frequency generators – these will be discussed in a separate eponymous article.

The autotransformer (AT) has two main purposes. The first is to supply voltage to the high-voltage transformer. The AT consists of a single winding on iron core. Special graphite slides (roles) are used to adjust the input line voltage and 

\[ kV - kVp \].

The secondary winding is usually with a grounded node 0.5–1.5 kV selector (F). The AT is placed either in a separate cabinet or in the control console (where all adjusting/command knobs are).

The high-voltage circuit includes also a timer, which interrupts the supply to the HVT after a given period of time, thus controlling the length of the x-ray exposure (ms). There are various designs for timers – usually electro-mechanic ones (old equipment) and electronic timers.

The filament circuit supplies and controls the filament current through the cathode of the x-ray tube. It includes the filament transformer (FT) – a step-down transformer with ratio of 1:10 and 1:20, which generates filament current of the order of 3–5 A through the cathode wire. This circuit could also include sub-circuits that have control of the space charge effect; control of the anode temperature influence over cathode wire temperature, etc. Often this circuit includes special voltage stabiliser.

The filament circuit controls the input to the FT through variable resistor (this way controlling the filament current, hence the anode current – mA). Additionally, it can include filament current adjustment resistors (F-select resistors, set by the service engineer). With these, the maximal filament current is limited to a set value. These adjustments are made with different KV/MA settings using the so-called three-point technique to properly select the necessary resistors (for: low KV@ high MA; high KV@ low MA; low KV@ low MA).

**FIGURE H.25** Block diagram of classical high-voltage x-ray generator.
Other special types of conventional HVG are capacitor-discharge generators, pulse-less (direct current) generators and monoblock generators. These will be discussed separately in eponymous articles.

Although the conventional HVG are still produced and used, a new type of design was introduced in the 1980s – the high-frequency high-voltage generator (or more correctly medium-frequency high-voltage generator). This HVG uses electronic converters which increase the electrical frequency to several kHz. In this new design the high-voltage transformer has ferrite core that is about 25% of the size of the traditional generator. All circuits are different and one can control the high-voltage by changing the frequency. For these HVG see the article High-frequency generators.

Related Articles: High-voltage generator, High-voltage transformer, High-frequency generator, Capacitor-discharge generator, Pulse-less generator, Monoblock generator, Filament circuit, High-voltage circuit, Voltage waveform

High-voltage protection
(Diagnostic Radiology) The high-voltage cables, the x-ray tube and all parts of the high-voltage generator (HVG) require special high-voltage (HV) insulation and protection. The insulation of the cables must withstand more than the maximal high-voltage tension of the equipment (e.g. 150 kV). The high-voltage transformer, the rectifiers, and all other HV parts are immersed in special insulating transformer oil, normally withstand- ing above 220 kV/cm. The same HV insulation is used inside the x-ray tube housing. Some x-ray devices have special system testing the HV protection and related electrical leakage (due mainly to internal capacitance).

Related Articles: High-voltage generator, X-ray tube housing

High-voltage transformer
(Diagnostic Radiology) The high-voltage transformer (HVT) is one of the most important parts of the high-voltage generator. It transforms the main voltage to the kV supplying the anode–cathode high voltage. It is part of the high-voltage circuit.

HVT is a step-up transformer with fixed transformation ratio of the order of 500–600. Its primary winding is connected to the mains and its secondary supplies the x-ray tube with high voltage (normally between 20 and 150 kVp). The secondary winding is usually with a grounded middle point (0 potential). Hence the HVT provides two halves of high voltage (for the maximal value of 150 kVp – these are +75 kVp to the anode and −75 kVp to the cathode), thus the electrical safety requirements (insolation) are more easily satisfied. The transformer is immersed in insulation transformer oil (normally more than 220 kV/cm insulation).

In the classical high-voltage generator this transformer is with iron core and is very massive (for power x-ray equipment it weighs hundreds of kilograms). There are significant ‘magnetic losses’ in its big laminated iron core, which leads to some delay of the exposure, as well as distortion of the porches of the x-ray pulse. In order to diminish this effect some HVT have special circuitry for constant pre-magnetising.

Power x-ray equipment using three-phase mains supply use three-phase high-voltage transformer, which has different types of secondary winding connection. Normally these are star-type connections for 6 pulse generator or star-delta connection for 12 pulse generator. See the diagrams in the article on Three phase generator.

The new high-frequency high-voltage generators (medium-frequency high-voltage generators) use high-voltage transformer with ferrite core. This allows use of higher frequencies, which minimise the size of the ferrite core down to 25% compared with iron core.

Related Articles: High-voltage generator, High-frequency generator, High-voltage circuit

HIS (Hospital information systems)
(Diagnostic Radiology) See Hospital information systems (HIS)

Histogram
(General) A histogram is a graphical representation of tabular data where one axis, typically the x-axis represents a domain divided into intervals. The value in each interval is denoted not by the height of the bar but by the area under the bar which in turn is determined by the number of observations in that specific interval. The total area under all the bars equals the total number of observations made (Figure H.26).

The histogram is considered to be one of the seven quality control (QC) tools.

Abbreviation: QC = Quality control.

HL-7
(Diagnostic Radiology) HL-7 is an accredited Standards Developing Organisation operating in the healthcare arena which aims to develop international healthcare standards. One of those standards is also called HL-7 and is a protocol for formatting, transmitting and receiving data in a healthcare environment. The HL-7 protocol development was started in 1987 with an aim to create a common ‘language’ allowing healthcare applications to share clinical data with each other.

The current version is HL-7 version 3 which was published in 2005.

The name is derived from the seventh and top layer, the application layer, of the ISO communications model for Open Systems Interconnection since the protocol is independent of all other, lower, communication layers.

Hogstrom algorithm
(Radiotherapy) The Hogstrom algorithm is an implementation of a pencil beam calculation for electron beams based on the Fermi–Egyes theory. In this theory, a narrow beam of electrons penetrating a medium is characterised by an increased lateral spread with depth due to multiple scattering processes. The Fermi–Egyes theory assumes small angle scattering, under which conditions the spatial distribution of the pencil beam is very nearly Gaussian at all depths. This Gaussian distribution can be characterised by the mean square of the scattering angle, mean radius-angle covariance and the mean square of the radius.

![Example of a histogram.](image)
A broad beam of electrons can be represented by summing the distribution for many smaller pencil beams (Lillicrap et al. 1975). In this case the dose to a point can be calculated by summing the contributions from all the pencils.

A heterogeneous medium, such as a patient, is represented by a stack of slabs, where heterogeneities along the central axis of the pencil beam are extended laterally to a semi-infinite homogeneous slab.

The assumptions of slab geometry and small angle scatter create some shortcomings in the approach (Figure H.27). Many of these were improved upon following the initial implementation, but more recently Monte Carlo calculations for electron beam dose distributions have become dominant.


![Generalised electron pencil beam approach. The irregular field is decomposed into elements. A ray trace is performed to a point of interest assuming slab geometry. The dose at a point is calculated from a summation of the pencil beams.](image)

**Holmium (General)**

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<th>Value</th>
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<td>Ho</td>
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<tr>
<td>Element category</td>
<td>Lanthanides</td>
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<tr>
<td>Mass number A of stable isotope</td>
<td>165 (100%)</td>
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</tr>
<tr>
<td>Density near room temp.</td>
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</tr>
</tbody>
</table>

**History:** Holmium was discovered in 1878 by Marc Delafontaine and Jacques-Louis Soret. It is a rare metallic earth element with a malleable form and high magnetic moment.

**Medical Applications**

Heavy Metal Filters in X-ray Imaging: Holmium is commonly used as a filter in the production of high quality diagnostic x-ray images where contrast agent (usually iodine or barium) has been employed. In diagnostic radiology maximum contrast is obtained when the incoming x-ray energy is close to, but slightly above, the K-edge of the contrast agent in question. For iodine the K-edge is 33.17 keV, and for barium the K-edge is 34.45 keV. The K-edge of a standard aluminium filter is 1.6 keV, whereas the K-edge of a holmium filter is 55.6 keV. Thus a holmium heavy metal filter will transmit a significantly narrower spectrum of photon energies than an aluminium one. The resultant reduction in the fluence of low energy photons will reduce the patient’s dose, and the reduction in the fluence of high energy photons will increase the image contrast.

Medical Lasers: As its laser light can readily be transmitted through optical fibres, holmium has found use as an active component in some medical (holmium-YAG) lasers.

Related Articles: Heavy-metal filter, Contrast media, K-edge metal filter

**Homogeneity**

(Magnetic Resonance) In MRI, the homogeneity of the main magnetic field governs the consistency of resonant frequency among the protons in a sample, such that good homogeneity is essential if slice selection, phase encoding and frequency encoding are to provide correct information from the patient.

One factor which affects magnetic field homogeneity is magnet design: solenoidal magnets with large bores generally have better uniformity than short bore magnets or open systems. In a given system, inhomogeneities in the static magnetic field may be caused by imperfect coil winding (variations in current densities within the wire), the presence of metal in the local environment or even the patient themselves. The process of making the magnetic field homogeneous is known as shimming.

So that the measure is independent of field strength, homogeneity is usually expressed in parts per million (ppm) relative to the main magnetic field. The level of homogeneity required for routine spin echo imaging is approximately 10 ppm, but for spectroscopy (where we are concerned with tiny differences in resonant frequency between metabolites) the minimum homogeneity increases by a factor of 100–0.1 ppm.

Related Articles: Shimming, Image homogeneity, RF homogeneity, B₀ homogeneity

**Radiotherapy** Homogeneity in radiotherapy usually refers to the degree of variation of dose within a given volume. Radiotherapy treatment planning and delivery usually strives to produce and deliver a uniform dose to the target volume, in other words the homogeneous dose to the target. Recognising that it is not always practical to get full homogeneity of dose in a target, ICRU (1993) has recommended that the degree of homogeneity should be kept within +7% and −5% of the prescribed dose.

Abbreviation: ICRU = International Commission on Radiation Units and Measurements.

**Homogeneity coefficient (HC)**

(related to Radiotherapy) Homogeneity coefficient, $h$, is the ratio of the first half value layer (HVL$_1$) to the second half value layer (HVL$_2$):

$$h = \frac{\text{HVL}_1}{\text{HVL}_2}$$

The first half value layer (HVL$_1$) is defined as the thickness of the specified material (absorber), which attenuates the air kerma or air kerma rate in the beam to one half of its original value, measured without any absorber. The contribution of all scattered radiation, other than any which might be present initially in the beam, is deemed to be excluded.

The second half value layer (HVL$_2$) is equal to the difference between the thickness of an absorber necessary to reduce the air kerma or air kerma rate to one quarter, $d_{1/4}$, and the value of HVL$_1$:

$$\text{HVL}_2 = d_{1/4} - \text{HVL}_1$$

The value of $h$ gives a certain indication about the width of the x-ray spectrum. Its value lies between 0 and 1 with higher values indicating a narrower spectrum. Typical values of $h$ for beams used in diagnostic radiology are between 0.7 and 0.9.

**Abbreviations:** HC = Homogeneity coefficient and HVL = Half value layer.

**Related Article:** Half value layer (HVL)

**Further Reading:** IAEA. 2007. Dosimetry in diagnostic radiology, an international code of practice. TRS 457, International Atomic Energy Agency, Vienna, Austria.

**Homogeneous field**

(Magnetic Resonance) The two main types of field characterising a MRI system are the main static magnetic field ($B_0$) and the RF field ($B_1$). Both fields need to be homogeneous to provide a perfect MR image, but this is more important for the main magnetic field. The $B_0$ magnet is made as homogeneous as possible, especially at the isocentre, to ensure that no unnecessary artefacts occur.

The RF field homogeneity will vary with the interaction between the RF field and the object being imaged and the choice of RF coil used in the system. Volume coils, such as the bird cage coil, produce high $B_1$ homogeneity over most of the coil volume, giving excellent image uniformity. Surface coils are characterised by their inhomogeneity, producing a high SNR at the surface of the object which rapidly decreases with depth.

**Abbreviation:** RF = Radiofrequency.

**Related Articles:** $B_0$, Homogeneity, $B_1$, Homogeneity

**Hormesis**

(Radiation Protection) The current internationally accepted framework for radiation protection is based on a model of potential harm from exposure to ionising radiation called the Linear No-Threshold Model. This model suggests that at any level of received radiation dose, harm may be caused – i.e. a cancer may be induced. The risk of harm (stochastic effects) is proportionate to the dose received (i.e. linear with dose). This is described in Figure H.28:

However, more recently published evidence would suggest that this model is too simplistic and that for a number of reasons low-level exposure to ionising radiation may actually be beneficial. This hormetic effect (hormesis) appears in epidemiological evidence from mortality rates amongst British radiologists, survivors of Chernobyl, and studies in Taiwan. Figure H.29 describes the possible consequence of hormesis to our understanding of the dose-response curve for stochastic effects.

Possible reasons for such a response have been postulated. Firstly, it is recognised that life on earth has evolved whilst continuously exposed to ionising radiation for natural sources, both terrestrial and from space. Secondly, it is recognised that the genome of humans and other animals seem to include ‘anti-cancer’ genes – genes that if ‘switched on’ appear to protect from the effects of being exposed to carcinogens. Ionising radiation exposure at low levels seems to be a trigger to switch on these anti-cancer genes.

The International Commission for Radiological Protection has stated in introducing its latest (2007) recommendations [4] that although it is accepted that hormesis is probably real, such a dose-response curve could not be used as the basis for a framework for occupational, medical, or public radiation protection because of the difficulties in specifying where the boundary lies between radiation doses that are beneficial and those higher doses that are deleterious.

**Related Articles:** Linear no-threshold model, Radiobiological models, Stochastic effects


**Hospital Information Systems (HIS)**

(Diagnostic Radiology) A Hospital Information System (HIS) is an information system product specifically for use in a medical or healthcare environment. It may be in the form of one single integrated software product or be composed of a number of separate systems that function together to manage information related to
the general administrative, financial and clinical work and service
delivery of a hospital or healthcare organisation.

The HIS may include more functionally specific systems such as
a radiology information system (RIS) or laboratory information sys-
tem (LIS), or it may share data with, and receive data from, external
implementations of such systems. The interconnectivity and shar-
ing of information means that patient and event related data need
only be entered once so minimising issues with data entry errors,
mis-registration or duplication of patients or events.

The sharing of data requires the various independent information
systems, often developed by different organisations, to implement
recognised information communication standards such as HL-7 as
promoted by the IHE (Integrating the Healthcare Enterprise).

Hot and cold spots
(Radiotherapy) Generally when planning is done for a treatment
an irradiation technique is developed which provides a maximum
and uniform dose to the planning target volume and minimises
the dose to both the treated volume and the irradiated volume as
defined by the International Commission on Radiation Units and
Measurements (ICRU 50, 1993). However for many reasons in
practice it is difficult to achieve the ideal dose distribution and
some dose heterogeneity has to be accepted. ICRU 50 (1993) rec-
tends that this dose variation should be kept within +7% (hot
spot) and −5% (cold spot) of the prescribed dose. Cold spots are of
concern inside the planning target volume (PTV) and hot spots in
those regions where organs at risk are located.

Therefore with these limits in mind the planning objectives would
be to ensure that the dose anywhere within the PTV does not fall
below 95% of the prescribed dose. A hot spot represents a dose out-
side the PTV that receives a dose larger than 100% of the specified
PTV dose. Additionally, dose volume is also an issue and it is consid-
ered clinically meaningful if the hot spot exceeds 15 mm in volume.

Further Readings: ICRU 50, 1993. Prescribing, reporting
and recording photon beam therapy. Report 50, International
Commission on Radiation Units and Measurements, Washington,
DC; Williams, J. R. and D. Thwaites. 2000. Radiotherapy Physics

Hot spot
(Radiotherapy) See Hot and cold spots

Hounsfield number
(Diagnostic Radiology) See CT number

Hounsfield scale
(Diagnostic Radiology) Many CT scanners use the term Hounsfield
units (HU) to express CT numbers (values proportional to the linear
attenuation coefficient of the material). The standard scale of HU
(Hounsfield scale) ranges from approximately −1000 to +3000 HU.
For more detail see the article CT number.

Related Article: CT number

Housing
(Diagnostic Radiology) See X-ray tube housing

Hue
(Diagnostic Radiology) See HSL (hue, saturation, luminance)

Hue, saturation, luminance (HSL)
(General) The well known RGB (red, green, blue) colour system
is very useful for colour image display on a monitor. This additive
system (all colours are formed as the sum of various amounts of the
three basic RGB colours) is used in devices using emitting light. The
alternative to RGB – is the subtractive CMYK system (using the
three reversed colours – cyan magenta, yellow plus black) is used
in devices reflecting colour – as in colour printing on white paper.

The RGB system has several limitations in digital imaging for
several reasons. The non-linear distances between the coordinates
(colours), makes less efficient image compression. A system which
overcomes this problem and creates a result better suited for the
human visual system is the non-linear colour system of HSL (hue,
saturation, luminance). In this system the colour space is created by
hue (wavelength); saturation (amount of colour, what can be expressed
as hue + grey); luminance (intensity, or brightness). Each one of these
three can be distinguished separately by the human visual system.

The fact that the HSL system has one channel for brightness and
two channels for chrominance (colour) presents a convenient situa-
tion, as human vision is very sensitive to brightness and less sensitive
to chrominance. This way the brightness channel can be transmitted
fully (e.g. with 8 bits), and the other two channels can be represented
with 4 bits each, what makes 16 bits in total – i.e. saves 8 bits from the
24 bits colour and helps the compression of the image data.

An alternative colour system is YUV, where Y is the luma (lumi-
nance) and U and V are the chrominance values. Closely related to
this system is YCbCr (where Y is the luma).

All these colour systems have special relationships with the
RGB colours. For example, the Y value in the YCbCr includes sub-
stantial green component (as the human eye is most sensitive to
green) and is given with the expression:

\[ Y = 0.299 R + 0.587 G + 0.114 B \]

while the other two channels have less green respectively more blue
or red:

\[ Cb = -0.1687 R - 0.3313 G + 0.5 B + 128 \]
\[ Cr = 0.5 R - 0.4187 G - 0.0813 B + 128 \]

Hyperlink: http://en.wikipedia.org/wiki/HSL_and_HSV

Humidity correction factor
(Radiotherapy) The calibration factor for an ionisation chamber is
valid only for the reference conditions which apply to the calibra-
tion and are specified by the standard dosimetry laboratory. Any
departure from the reference conditions when using the ionisation
chamber in the user beam should be corrected by using appropriate
factors. The reading of a dosimeter is usually corrected for pres-
ture, temperature, humidity, polarity effect and ion recombination.

The values of \( W \), where \( W \) is the mean energy required to cre-
at an ion pair in air and \( e \) is the electron charge, and the value of
the stopping powers appearing in equation used to calculate the
absorbed dose from ionisation measurements apply to a dry air.
Therefore the ionisation measurements should be done with a very
low relative humidity. The value of \( W \) decreases with the humidity
and therefore at low humidity conditions the amount of ionisation
increases. On the other side the stopping powers increases with
humidity tending to decrease the charge collected by the ionisa-
tion chamber. The two opposite effects combine so that the correc-
tion factor for humidity is approximately constant for the range of
 humidities normally encountered by the user at room temperature.

A humidity calibration factor of 0.997 should be applied for an
ionisation chamber whose calibration factor refers to dry air, usu-
ally for calibrations in \(^{60}\text{Co}\) in terms of air kerma. The factor is
Huygens’ principle

No corrections for humidity are needed if the calibration factor refers to a relative humidity of 50% and is used in a relative humidity between 20% and 80%.

A high relative humidity can cause higher leakage currents in the ionisation chamber and in the electrometer. Therefore it is advisable to check them mainly when the ionisation chamber is built from hygroscopic materials such as A-150 plastic or nylon.

Hydrophone

A hydrophone is a device used for listening to sound in liquids, like a microphone does in air. They are made of a piezoelectric material that converts the pressure changes in the sound waves to electrical signals and the sensitivity of a hydrophone is expressed in V/Pa. Ideally they should be small, sensitive, linear with broad bandwidth. There are three different commonly used types of hydrophones: ceramic hydrophones, pvdf needle-probe hydrophones and membrane hydrophones.

The ceramic hydrophone is constructed like a small single element transducer with a disc-shaped sensor made of absorbing materials. Because of resonance and reflection in the material the accuracy will be poor. This can be prevented if the hydrophone is properly calibrated. Ceramic hydrophones are robust devices and are often used in physiotherapy equipment.

The needle-probe hydrophone, Figure H.30, is very similar to the ceramic hydrophone but is smaller (diameter 0.02–1 mm) and uses a piezoelectric polymer film as the sensor material, polyvinylidene fluoride (pvdf). The acoustic impedance of the pvdf-film is almost the same as for water and this is important to reduce frequency variations of sensitivity. So the needle-probe hydrophone is fairly frequency

Hydrogen

Symbol H
Element category Non-metal
Mass number A 1
Atomic number Z 1
Atomic weight 1.008 g/mol
Electronic configuration 1s1
Melting point 14.01 K
Boiling point 20.28 K
Density near room temperature 0.0899 g/L

History: Molecular hydrogen gas was first produced artificially by T. Von Hohenheim in the sixteenth century. It was only identified as a new element in experiments performed by Henry Cavendish between 1766 and 1781. The name hydrogen is derived from hydro, water, and genesis, to make, and was given to the element because of the water formed when hydrogen gas is burned.

Isotopes of Hydrogen: By far the most common isotope of hydrogen is protium, 1H, which is stable and forms 99.98% of naturally occurring hydrogen. Hydrogen also exists in trace amounts as stable 2H, deuterium, and radioactive tritium, 3H. As a whole, hydrogen makes up over 75% of the universe by mass. There also exist unstable isotopes of hydrogen from 4H to 7H, which do not occur naturally. Hydrogen is formed as a by-product of splitting hydrocarbons. It can also be produced by electrolysis at high cost.

Tritium decays to 3He through low energy beta particle emission, with a half-life of 12.32 years.

Medical Applications: NMR – 1H is the source of protons detected in most NMR investigations. Signal from the proton of 1H bound in biological tissue can be collected to obtain information about the chemical composition of the tissue, as in NMR spectroscopy, or to form an image of the hydrogen distribution and environment, as in MRI.

Deuterated water (D2O), or heavy water, is used instead of protium-based water as a solvent for NMR spectroscopy phantoms. Normal water protons would interfere with the signal from the chemical solutes used to obtain the spectrum. The 2H proton produces a signal at a different frequency, enabling it to be separated from the solute signals.

Biomedical Research Tracer – Tritium, 3H, is a common marker used to investigate physiological processes, particularly in biomedical research and less commonly as a tracer in clinical PET studies. Since hydrogen is found in most biological molecules, 3H can be bound to a wide range of metabolites and tissues. The electron produced in its decay has an average energy of 5.7 keV, so it cannot penetrate the body. Tritiated water is used as a base for many research products and drug safety studies.

Related Article: Magnetic resonance imaging (MRI)

FIGURE H.30 Needle-probe hydrophone. (Graphs courtesy of EMIT project, www.emerald2.eu)

FIGURE H.31 Membrane hydrophone. (Graphs courtesy of EMIT project, www.emerald2.eu)
stable but for lower frequencies a rapid fall-off of sensitivity due to diffusion around the probe tip. This type of hydrophone is commonly used to characterise ultrasonic fields from diagnostic equipments.

A membrane hydrophone, Figure H.31, is also using pvdF as sensor material. The polymer film is stretched over an annular ring of about 100 mm diameter. The electrodes on each side of the film only overlap in a tiny area (diameter 0.05–0.5 mm), which is poled and forms the active sensor element. The advantage with this type of hydrophone is the stable frequency response.

**Related Article:** Acoustic pressure

**Hyperechoic**

(Ultrasound) See Echogenic

**Hyperfractionation**

(Radiotherapy) Hyperfractionation refers to the use of a high number of fractions, with dose fractions smaller than the conventional 1.8–2 Gy per fraction, usually delivered more than once a day. Where this results in an overall treatment time less than that for conventional fractionation it is referred to as accelerated hyperfractionation.

Generally, radiotherapy treatment is delivered once a day, 5 days a week for up to 8 weeks. Each daily treatment is called a fraction and such regimens have been shown to be clinically effective and acceptable, giving a favourable therapeutic effect in most cases. Better tumour control is obtained for a given level of normal tissue toxicity when the radiation dose is fractionated rather than delivered as a single dose. The effectiveness of fractionated radiotherapy can be understood by consideration of the 5R's of radiobiology. Dividing the radiotherapy dose into fractions spares normal tissues since the cells can repair some of the radiation damage in the time between fractions and cell repopulation will occur provided the overall time is sufficiently long. Conversely, fractionating the dose increases the damage to the tumour due to reoxygenation and the redistribution of cells into radiosensitive phases of the cell cycle between fractions.

Although the prolongation of treatment by the use of fractionation is beneficial in many cases, the excessive prolongation of treatment may actually have a detrimental effect on the therapeutic efficacy of the treatment if proliferation of tumour cells becomes significant. Irradiation of tumour cells can trigger the surviving cells to divide faster than before. This is known as accelerated repopulation and further details can be found in the article on Repopulation. Hyperfractionation, particularly accelerated hyperfractionation, offers a possible solution to this problem.

The aim of hyperfractionated radiotherapy is to exploit the difference in the effect of the dose per fraction on the tumour and acute-responding normal tissues compared with late-responding normal tissues, thereby decreasing the incidence of late normal tissue effects whilst maintaining both the tumour control and the incidence of early effects of the conventional treatment. In clinical practice, this results in an increase in the total delivered dose and sometimes also an increase in the overall treatment time. Accelerated treatment aims to reduce repopulation in rapidly proliferating tumours by delivering the same total treatment dose as the conventional regimen but in a shortened overall time achieved by giving two or more fractions a day. In practice, it is not possible to achieve this due to the increased incidence of early normal tissue effects. Accelerated hyperfractionated treatment is a combination of the accelerated and hyperfractionated treatment protocols. A low total dose is delivered in a short overall treatment time by the delivery of multiple, small doses given at least 6 h apart. Continuous hyperfractionated accelerated radiation therapy (CHART), which reduces overall treatment from 6 to 7 weeks to 12 days and gives 36 small fractions, has been tested in multicentre randomised controlled clinical trials. The trial in non-small-cell lung cancer showed improvement in survival and this regimen is now the government recommended standard of care for eligible patients in the United Kingdom. It should be noted that early normal tissue effects were more severe than for a conventional regime but patients found them tolerable and they did not translate to an increase in late normal tissue toxicity.

In cases where a change in fractionation regimen is considered, perhaps due to an interruption of treatment or as part of a clinical trial, it is useful to be able to compare the regimens in terms of the effect on both the tumour and the normal tissues. This is possible by using the linear-quadratic model to evaluate the biological effective dose (BED). In the case of normal tissues, a more useful parameter may be the equivalent total dose in 2 Gy fractions (EQD2) since most clinical experience of normal tissue tolerance has been obtained from 2 Gy per fraction regimens. More detail can be found in the article on Biological effective dose.

**Abbreviations:** BED = Biological effective dose and EQD2 = Equivalent total dose in 2 Gy fractions

**Related Articles:** Adverse effects, Alpha–beta ratio, Biological effective dose, Cell cycle, Fractionation, Interruption of treatment, Linear quadratic (LQ) model, Radiosensitivity, Repair, Repopulation, Reoxygenation, Therapeutic effect, Tolerance, 5R's of radiobiology


**Hyperpolarised**

(Magnetic Resonance) The size of the longitudinal magnetisation available in an NMR experiment places a practical limit on MRI spatial and temporal resolution. In most cases, this magnetisation is generated by and evolves in the static field of the MRI magnet. Because the difference in populations of the two energy levels brought about by exposure to the static field is very small (a few nuclei per million), this magnetisation and the resulting signal are also correspondingly small. However, there are situations in which much larger magnetisation is created, either by transient exposure to a very strong polarising magnetic field or by other means. Although this magnetisation then evolves in a weaker field, far more signal is available because of the larger initial magnetisation. This technique is known as hyperpolarisation.

From the Boltzmann equation, it follows that exposure to a very strong field, combined if possible with cooling to very low temperature, would greatly increase the size of the available magnetisation. This approach has been exploited in systems that use a transient strong magnetic field to induce polarization, the resulting magnetisation then evolving in the Earth’s magnetic field.

However, in general, less direct methods of hyperpolarisation have been used, usually as a means to image the gases 3He and 129Xe. These methods involve using a laser to generate a large electron spin polarisation, either in the target atom itself or in some intermediary such as rubidium, which is then transferred to the nuclei of interest via hyperfine interactions within the atom, or via collisions. These
Hypofractionation

(Radiotherapy) Hypofractionation refers to the use of a lower number of fractions, with dose fractions substantially larger than the conventional 2 Gy per fraction.

Hypofractionation and single high-dose irradiation are widely used in palliative radiotherapy with doses typically ranging from 3 to 10 Gy. Lower total doses are delivered than for radical treatment and this combined with the limited life expectancy of the patients means that late normal tissue damage is not a major concern. A number of randomised clinical trials of such regimens have demonstrated equivocal symptom control to more fractionated schedules and have the advantage of being more convenient for the patient.

Recently, interest in hypofractionation for radical treatment has renewed in the light of new data on the alpha–beta ratio values for some tumours, the results of fractionation trials such as START, and the advent of intensity modulated radiation therapy. Many of the hypofractionated regimens currently in use for radical radiotherapy have developed as a result of expediency rather than from radiobiological principles. Historically, the most common dose per fraction in use has been 2 Gy. However, in the northern United Kingdom fraction sizes of 2.67–2.75 Gy per day have been used for many years, introduced largely to ease the burden on thinly spread resources.

The moderate hypofractionated regimes used routinely in some centres in the north of the United Kingdom are delivered with slightly lower total doses applied than those for conventional 2 Gy per fraction regimens (e.g. for carcinoma of the prostate 55 Gy in 20 fractions has been compared with 66 Gy in 33 fractions) to reduce the risk of late normal tissue damage. For tumours with a high alpha–beta ratio this may lead to a reduction in tumour control probability, although some of the negative effect may be compensated for by the reduced overall treatment time. However, for tumours with low alpha–beta ratios, in the range of 2–3 Gy and therefore similar to that for late responding tissues, delivery of a smaller number of larger dose fractions should result in good local tumour control without an increase in normal tissue toxicity. In the case of prostate cancer, evidence is accumulating that its alpha–beta ratio is in the low region.

In the United Kingdom, the CHHIP randomised controlled trial will compare 2 Gy dose per fraction (total dose 74 Gy) and 3 Gy dose per fraction (total dose 57 and 60 Gy). In breast cancer, the START B trial compared 50 Gy delivered in 20 fractions to 40 Gy delivered in 15 fractions and found no difference in local tumour control between the two arms but lower rates of late adverse effects were observed in the hypofractionated arm. Analysis of the data from the START A trial indicated an alpha–beta ratio for breast cancer of around 4 Gy.

Intensity modulated radiation therapy (IMRT) and other highly conformal techniques such as tomotherapy and proton therapy, potentially deliver improved dose distributions over conventional radiotherapy with lower normal tissue doses for comparable tumour doses. This may mean it is possible to increase the dose per fraction delivered to the tumour since the lower dose per fraction delivered to late responding normal tissues reduces the need to spare them. The delivery of a simultaneous integrated boost (SIB), whereby different doses per fraction are delivered to different target regions, is gaining popularity and is incorporated into a number of IMRT clinical trials.

In cases where a change in fractionation regimen is considered, perhaps due to an interruption of treatment or as part of a clinical trial, it is useful to be able to compare the regimens in terms of the effect on both the tumour and the normal tissues. This is possible by using the linear-quadratic model to evaluate the biological effective dose (BED). In the case of normal tissues, a more useful parameter may be the equivalent total dose in 2 Gy fractions (EQD2) since most clinical experience of normal tissue tolerance has been obtained from 2 Gy per fraction regimens. More detail can be found in the article on Biological effective dose.

Related Articles: Boltzmann distribution, Longitudinal magnetisation, Static field

Hz (hertz)

(General) See Hertz (Hz)
IAEA  
(General) See International Atomic Energy Agency (IAEA)

ICNIRP  
(Radiation Protection) See International Commission on Non-Ionising Radiation

ICRP (International Commission on Radiological Protection)  
(Radiation Protection) See International Commission on Radiological Protection (ICRP)

ICRU  
(Radiation Protection) See International Commission on Radiation Units and Measurements

ICRU reference point  
(Radiotherapy, Brachytherapy)

ICRU Report 38 – Reference Points for Reporting Doses to Organs at Risk: The ICRU reference points for bladder and rectum are described in the ICRU Report 38 ‘Dose and Volume Specification for Reporting Intracavitary Gynaecology’. The reference points are defined in relation to the applicator and the doses to these points are calculated from orthogonal radiographs.

A Foley catheter, with the balloon filled with 7 cm³ contrast fluid, is used to define the bladder reference point. On the lateral film this point is defined as the posterior surface of the Foley balloon, on the vertical AP axis drawn through the centre of the balloon. On the frontal (AP) film, the balloon centre point is used.

On the lateral film, the rectal reference point is located on a vertical line drawn from the lower end of the intrauterine source. The point is defined 0.5 cm posterior to the vaginal wall, which is visualised, e.g. using a radio-opaque gauze packing. On the frontal (AP) film the reference point is located at the lower end of the intrauterine sources or at the midpoint between the vaginal sources.

ICRU Report 38 – Reference Points for Specification of an Intracavitary Application: The ICRU Report 38 recommends the use of ‘reference volume’ together with the total reference air kerma for the application, instead of dose to a reference point in a region with a high dose gradient.

Nevertheless, two points called A points, defined in relation to the applicator as ‘2 cm up and 2 cm out laterally left and right’ have been used extensively both for specification and reporting (point A was originally defined in the Manchester system).

Reference Points – Reporting for Image-Guided Techniques: Modern recommendations for reporting intracavitary brachytherapy are based on image-guided techniques and dose-volume histogram parameters. But, both the ICRU reference points and point A are still included for reporting purposes, to relate the newer image-guided techniques to the older techniques.

Abbreviations: AP = Anterior – posterior and ICRU = International Commission on Radiation Units and Measurements.

Related Articles: Internal reference point, Reference volume, Manchester system

**CT Artefacts:** In the case of CT the presence of high Z implants such as artificial hips or implanted gold fiducial markers lead to streaking (see Figure I.1). The effects of patient motion, such as that due to breathing, may lead to distortion of the target shape and organ shapes if the CT scanner is not synchronised with the breathing motion using 4D CT.

**MRI Artefacts:** MRI is also prone to artefacts caused by patient movement during scanning. In addition to this spatial distortion occurs due to variations in magnetic field strength and magnetic susceptibility effects.

**Abbreviations:** CT = Computed tomography, 4D CT = Four dimensional computed tomography and MRI = Magnetic resonance imaging.


**Image compression**

(General) Image compression is a digital image processing technique which attempts to reduce the amount of data needed to represent the information in an image.

Many compression techniques exist which can reduce the original image pixel values into a much smaller coded block of data. The techniques may be divided into ‘lossy’ and ‘non-lossy’ processes.

Non-lossy compression is able protect all the original image data and it is possible to exactly reconstruct the original image without loss of any information. This is therefore favoured in medical applications where images are used for diagnosis, though the amount of compression available (size reduction) may only be within one order of magnitude.

Lossy compression techniques are now common where image data need to be transmitted quickly, and some amount of image loss can be accepted. Common image storage formats such as JPEG and MPEG include compression, though the amount of compression and the restriction to lossless compression is usually selectable. Lossy compression techniques may reduce image data size by large factors (1/10–1/100) with only minimal reduction in observed image quality.

**Image covariance**

(General) A mathematical measure of the similarity of two digital images.

**Image display**

(General) In medical imaging, digital images are displayed on cathode ray tubes (CRT) or liquid crystal displays (LCDs).

Individual pixels in an image are displayed with different brightness levels depending on the pixel value. The images can be displayed in black and white (greyscale) where the levels are represented by varying degrees of white light or colour where the levels are assigned to a colour scale consisting of varying degrees of red, green and blue light (RGB).

Image display systems are characterised by a number of parameters, examples of which are spatial resolution, spatial distortion and contrast resolution. In order to preserve image detail, the spatial resolution of a display device should be greater than that of the underlying image.

**Related Articles:** Grey levels, RGB

**Image enhancement**

(Ultrasound) Image enhancement is a general term covering adjustments to an image in order to improve or alter the image and its appearance. In medical imaging it encompasses a range of techniques to improve the detection of features in images. Techniques include:

- Post-processing, ascribing different grey levels or colours to the final output
- Adaptive processing (Figure I.2)
- Edge enhancement
- Temporal filtering, e.g. persistence

Adaptive processing produces smoothing in the ultrasound image of a phantom with more consistent contrast in the cylinder targets. The processing reduces background speckle in the phantom gel material but introduces patterns in the image that are not in the original.

**Related Articles:** Post-processing, Adaptive processing, Edge enhancement, Persistence

**Image fusion**

(Nuclear Medicine) Nuclear medicine and PET images suffer greatly from a lack of anatomical detail. In order to provide anatomical structure, computer algorithms have been developed to accurately align these images with images taken from CT or MRI. It is very useful to display the anatomical detail (e.g. provided by CT) and the physiological detail (e.g. provided by SPECT) on the same image, one overlying the other. This is known as image fusion.

**Related Article:** Registration

**FIGURE I.1** CT scan of prostate with implanted markers. Note the streak artefacts extending diagonally across the centre.

**FIGURE I.2** Adaptive processing.
Several forms of image distortion can occur in projection x-ray imaging and are related to the geometry of the process. Three types of geometric distortion are illustrated later (Figures I.3 through I.5).

Other geometric distortions exist in fluoroscopy. The main such image intensifier (II) distortions are as follows:

- **Pincushion distortion** – it is caused by a non-uniform magnification across the image, with magnification increasing away from the image centre. This is due to the fact that the peripheral photo electrons from the II photocathode are less affected by the accelerating electro-magnetic field. Usually pincushion distortion is prominent at larger II fields (see the eponymous article).

- **Barrel distortion** – an opposite to pincushion distortion. This effect can be due to various factors, one of which is pre-compensation of the Pincushion distortion (see the eponymous article).

- **S distortion** – it is mainly due to external electromagnetic fields. As a result the peripheral photo electrons from the II photocathode are under the influence of both the accelerating electro-magnetic field and the external field. This combined influence creates S distortion (see the image in the article on Vignetting).

**Related Articles:** Pincushion distortion, Barrel distortion, Vignetting

**Image intensifier**

*Diagnostic Radiology* The image intensifier (II) is an electronic vacuum device used to intensify electronically the brightness of the x-ray image. II is the main detector used in x-ray fluoroscopy. The dose necessary to produce an image with the II is significantly less than the dose necessary to produce a film radiograph.

Typical block diagram of an II is shown on Figure I.6. The envelope (4) of the II is made by glass (in older II types) or mu-metal (newer II). In the case of mu metal the Input window (also called entrance window) 1 is made of very thin metal (as 0.2mm aluminium) which assures 95% transmission of the falling x-ray radiation (already modulated by the absorption of the tissues of the patient). The x-ray quanta passing through 1 hit the input phosphor (2) and produce light with intensity proportional to the x-ray beam intensity. Most of the new II use caesium iodide (often CsI:Na) as input phosphor. This material emits light with spectrum matching closely the sensitivity of the next layer – the photocathode 3 (usually made of antimony and caesium). The light photons from the input phosphor interact with the photocathode, which produces photoelectrons. The number of photoelectrons produced is proportional to the number of light photons, hence to the intensity of the x-ray beam. A system of accelerating electrodes (5) focuses and accelerates the photoelectrons to approximately 30keV. These accelerated photoelectrons hit the output phosphor 7 and produce a small but very bright light. Most II use zinc sulphide (often ZnCdS:Ag) as output phosphor. This bright image from the output phosphor passes through a glass window 8 and through fibre optic or optical system (9) is transferred to the TV camera tube (and from the to the diagnostic TV monitor). A thin (approx. 0.2μm) non-transparent metal anode (6) assures that the bright light from the output phosphor does not light back the photocathode.

Usually the electric potential of the photocathode is 0V, and the potential of the accelerating electrodes increases reaching +30keV at the anode in front of the output phosphor. The equipotential

![Image intensifier block diagram](image.png)
lines of the accelerating electric field form an ‘electric lens’ which focuses precisely the beam of photoelectrons (the dashed lines on Figure 1.6). Any external electromagnetic fields can distort the ‘electric lens’ and hence the image. Due to this reason the II has to be shielded (hence the need of mu-metal for its envelope).

Figure 1.7 shows an old II with glass envelope. The front phosphor is clearly seen. The diameter of this II is 23 cm (compare with the pencil on the photo). Usually II are made with standard front diameters such as 15, 23, 30, 40 cm (in inches from 6 to 16 in. diameter). Figure 1.8 shows the same II with broken entrance window. The image clearly shows the cylindrical accelerating electrodes and the small output phosphor at the back (usually with diameter of 2.5 cm).

There are a number of parameters used to assess an image intensifier (II). The most important being:

- Total brightness gain = (output light photons)/(input x-ray photons)

Usually this figure is between 1000 and 6000.

- Conversion factor = (output phosphor light)/(input screen dose rate)

Usually this figure is between 100 and 1000 (cd/m²/µGy/s).

The tandem optics (with semitransparent mirror) at the output of the II allows splitting the light beam to two – one to the TV camera and monitor, the other to some recording device as cine camera, spot camera, etc. (see related articles).

The automatic brightness control (ABC) allows monitoring the dose rate at the entrance window of the II by inserting a probe which measures the intensity of the light exiting the II in the parallel light beam formed at the optics level (9).

The newest flat panel detectors used in digital radiography can work in fluoroscopic mode as well. It is expected that in future these will replace the image intensifiers.

**Related Articles:** Fluoroscopy, Total brightness gain, Cinelluoroscopy, Spot camera

**Image noise**

(Diagnostic Radiology) Visual noise in an image is a generally undesirable characteristic of the image that reduces visibility of low contrast objects (Figure 1.9). In x-ray and radionuclide imaging the statistically random distribution of photons received by the receptor is the predominant source of noise.

In general image noise is the stochastic variation of the image signal. Quantitatively it is described by the SNR (signal to noise ratio) or NPS (noise power spectrum) of the system.

There are various sources of noise – the imaging radiation, the detector and system hardware, the reconstruction software, etc. A most often encountered noise in imaging systems using ionising radiation is quantum noise, which is due to the stochastic variation of x-rays, photons, etc.

In digital systems noise is often estimated by the standard deviation of the pixel values within certain region of interest. Figure 1.10 gives an example of image noise (the visual granularity) in a CT scan of bottle with water, which should represents an uniform image with CT number 0 (in the image is 0.4), but in reality the water is presented with a granular structure of different pixel values (St. Dev. 6.62). This representation of the noise in a CT image is related to various factors, which for CT are described by the formula of Brooks and Di Chiro:

\[ \sigma \sim C \left( \frac{e^{-md}}{D*ST*P} \right)^{1/2} \]

**Effect of noise on visibility**

Object contrast

Low

High

Increasing noise
which shows that the CT noise (μ) depends mainly on

- Reconstruction algorithm and filter – C
- Object parameters – (μd)
- Scan dose – D
- Slice thickness – ST
- Pixel size – P

Related Articles: Contrast resolution, Signal to noise ratio, Noise power spectrum

Image noise
(Magnetic Resonance) Image noise is unavoidable in MRI, as in all other imaging modalities. There are two main sources of noise in an MR image: electromagnetic noise in the body of the patient due to the movement of charged particles within human tissue, and Johnson noise in the RF coil and associated electronics within the MRI system. Under most circumstances, patient-generated noise is dominant. However, when imaging at low field, or when using small receiver coils even at 3T, noise generated in the coil can be more significant.

A number of parameters affect the amount of noise within an image including bandwidth, matrix size, slice width and acquisition timing parameters such as TR, TE and TI. The choice of RF coil(s) will also have an impact, with a small tightly coupled coil generally demonstrating less noise than a larger coil. The amount that this contributes to the noise depends on the size of the RF coil and the bandwidth of the pulse sequence. A large RF coil or a wide bandwidth will increase noise.

Noise is normally quantified by measuring the signal to noise ratio (SNR). This is a more useful number than considering noise alone, since the strength of the signal will determine how much noise will be acceptable in the image before it ceases to be diagnostically useful. Random noise gives the MR image a slightly mottled effect and the lower the SNR, the more mottled the image becomes.

As well as noise generated within the imaging process itself, MR images are susceptible to external RF noise from a range of sources, such as radio and television broadcast, noisy electronic components in other pieces of equipment, and static electrical discharge. These may also contribute random noise, but in some cases the noise may be structured, for example the ‘herringbone’ artefact due to electrical spikes and discrete bands of noise due to narrow-band RF transmissions.

Abbreviations: SNR = Signal to noise ratio.

Image noise
(Nuclear Medicine) Image noise is a common feature in diagnostic imaging, particularly in nuclear medicine where the count rates (or information densities) are low. Noise can be divided into two categories depending on their nature and behaviour. The first category is called random noise, which is basically statistical noise present in all images as a mottle background. Random noise is caused by several factors, e.g. statistical variations in counting rate and static noise in circuitry.

The second type is structured noise and it refers to nonrandom effects on the counting rate. Such non-random variations can be caused by uptake in superficial organs blocking the view of a deeper lying organ of interest. Other kinds of structural noise are image system artefacts, e.g. non-uniformities in gamma camera or image reconstruction artefacts.


Image perception
(General) Image perception is the process by which the information in images is viewed and interpreted. In medical imaging, the subject encompasses the techniques of image acquisition, processing and display and also human perceptual and interpretation processes.

Image plane
(Magnetic Resonance) MRI is inherently a three-dimensional imaging technique, with the additional step of slice selection needed to restrict data acquisition to a slice rather than a volume of tissue. Nevertheless, in most applications images are collected from a stack of slices. The orientation of these slices is known as the image plane.

In MRI the image plane is determined by the orientation of the slice selection gradient. It can lie in any one of the principal anatomical planes (axial, coronal or sagittal), or by combining two or three physical gradients images can be collected in any oblique plane. This extremely useful capability allows images to be collected, for example in the long and short axes of the heart, which are not conveniently aligned with either anatomical planes in the patient or the Cartesian axes of the MRI scanner.

Related Articles: B₀ gradients, Multiplanar reconstruction, Oblique imaging, Slice selection

Image processing software
(General) Medical imaging systems include built-in image processing capability provided by the modality vendor. This capability usually resides on the acquisition console of the modality or on an associated satellite computer system. Reporting stations on PACS systems will also have image processing capabilities. Image processing applications running on clinical systems for day to day use guide the user via graphical user interfaces (GUIs) and do not require any programming skills. For implementation of more sophisticated techniques or for the development of new techniques, off line image processing using dedicated software may be necessary. Several image processing suites appropriate for medical
Spatial Resolution: Spatial resolution defines the smallest detectable detail of an image. The smaller the voxels of an image, the higher is the resulting spatial resolution. The volume of a voxel depends on the image matrix size (e.g. 256 × 256), the field of view (FOV) and the slice thickness. Because only a finite part of k-space is covered during the imaging process, and various other physical limitations, the actual spatial resolution is usually poorer than the theoretical pixel limit.

Signal to Noise Ratio: The signal to noise ratio (SNR) is used to define the relative contributions to an image of the detected signal and random noise. Several different definitions of SNR are in use, but a common one is the ratio of the average signal intensity within a region of interest (ROI) to the standard deviation of the noise within that ROI. SNR can be improved by using a detection coil optimised to the part of the body being imaged, to prevent the detection of noise from adjacent structures, by summing several signal measurements, by sampling larger volumes (reducing spatial resolution by increasing the FOV and slice thickness) and by optimising the receiver bandwidth.

Image Contrast and Contrast to Noise Ratio: The contrast in an MR image depends on the type of pulse sequence used and its parameters. Structures with little contrast between them are difficult to distinguish in the presence of noise, and this can be expressed in the contrast to noise ratio (CNR), the ratio of the signal difference between two objects to the noise present in the image.

Artifacts: Artifacts are features of the image that are not found in the imaged object, but are introduced due to imperfections in the imaging process or to confounding factors such as patient movement. It is important to be familiar with the appearance of artifacts because they can be mistaken for pathology. Numerous kinds of artifacts can occur in MRI and a few of them are summarised in Table I.1.


<table>
<thead>
<tr>
<th>TABLE I.1</th>
<th>A Summary of Artefacts in MRI, Their Causes and Appearance</th>
</tr>
</thead>
<tbody>
<tr>
<td>Artefact Type</td>
<td>Cause</td>
</tr>
<tr>
<td>Motion</td>
<td>Random or periodic movement of the imaged tissue</td>
</tr>
<tr>
<td>Susceptibility</td>
<td>Difference in susceptibility between adjacent tissues or between tissues and air</td>
</tr>
<tr>
<td>Metal</td>
<td>Difference in susceptibility between metal implants/foreign bodies and surrounding tissue</td>
</tr>
<tr>
<td>Flow</td>
<td>Movement of blood through imaged region</td>
</tr>
<tr>
<td>Chemical shift</td>
<td>Difference in Larmor frequency between fat and water</td>
</tr>
<tr>
<td>Gibbs</td>
<td>Insufficient sampling of k-space</td>
</tr>
<tr>
<td>Aliasing</td>
<td>Inadequate FOV size, so Nyquist criterion is not satisfied</td>
</tr>
</tbody>
</table>

Image quality

(Magnetic Resonance)

Definition: Image quality is a general term used to describe how faithfully an image reproduces features of the imaged object and the degree of visibility of relevant information in an image. Good image quality is essential to achieve diagnostic quality in MRI and other imaging modalities. It is often assessed using quality control protocols provided by the manufacturer or by independent sources (e.g. EUROSPIN). MR image quality is characterised by several factors, including spatial resolution, image contrast, signal to noise ratio and the presence of image artefacts.
Image reconstruction

(Magnetic Resonance) This term relates to the process of converting raw data that have been acquired using an MRI scanner into an image.

In conventional Fourier imaging, raw data take the form of a two-dimensional array of complex data in k-space. Each item of data in the array represents the vector sum of magnetisation throughout the excited slice following a specific phase encoding episode and at a specific point during the application of the frequency encoding gradient. Each data point therefore contains information about the entire image, but the phase information encoding position in these data points cannot be extracted directly.

The signal element at a given location in k-space is given by

\[ S(k_x, k_y) = \int \int \rho(x, y, z) e^{-2\pi i (k_x x + k_y y)} \, dx \, dy \, dz \]

where \( \rho(x, y, z) \) is proton density – essentially the spatial distribution of water protons that we wish to recover (in general, weighted by relaxation times and other parameters, depending on the pulse sequence). It can be seen that two-dimensional Fourier transformation of the entire array will yield the desired information.

In some applications, non-cartesian k-space trajectories are used, and these require special treatment in terms of reconstruction. In radial imaging, a series of radial lines through k-space are acquired by applying two orthogonal gradients simultaneously, the relative amplitude of the two gradients determining the angle in k-space of a particular acquisition. The image can be reconstructed by backprojection, or more commonly by interpolating the radially-acquired data onto a Cartesian grid and performing two-dimensional Fourier transformation as usual.

Similarly, data acquired using a spiral k-space trajectory must undergo Cartesian regridding prior to Fourier transformation. In partially parallel imaging, the redundancy inherent in data collected using multiple receiver elements is exploited, either in k-space or in image space, to reduce the required number of phase encoding steps (and hence image acquisition time) without sacrificing spatial resolution or field of view (FOV). Image reconstruction using these techniques requires additional mathematical steps either to fill in omitted parts of k-space prior to Fourier transformation or to recover useable images from severely aliased images with restricted FOVs afterwards.

Related Articles: k-space, PPI (partial parallel imaging), Raw data

Image reconstruction

(Nuclear Medicine) This is the process of using acquired 2-D projections to build a 3D image volume. The reconstruction procedures are more elaborately described in their separate articles. Image reconstruction is used in all tomographic imaging (CT, MR, PET and SPECT). The goal is to achieve a reconstructed image volume with a high contrast and high signal to noise ratio (SNR). These two parameters are not only affected by the choice of reconstruction method, but by a number of factors, such as acquisition time, administered activity, etc. One of the earliest methods suggested was filtered back projection. The image data are transformed into Fourier space and filtered, then transformed back to the spatial dimension where it is reconstructed according to the back projection theorem. This reconstruction method produces an adequate result with low computational power required.

Acquired data do not only consist of ‘true’ events but are also ‘contaminated’ with false events due to a number of reasons, e.g. scattered radiation and noise from electric circuitry. These effects can be suppressed in the reconstruction if the effect can be modelled. One example of a reconstruction method where it is possible to remove false events is called Iterative reconstruction.

An iterative reconstruction is started with a ‘guess’ or an estimation of the acquired image. Projections from this estimated image are derived and compared to the actual measured projections. Using information from this comparison the estimated image is updated and the procedure is repeated. Iterative reconstruction was considered to be to computationally heavy but recent advances in computer technology has lead to a much greater use of this method. Two different types of iterative reconstruction are OSEM (ordered subset expectation maximisation) and MLEM (maximum likelihood expectation maximisation).

Related Articles: Iterative reconstruction methods, Filtered back projection, Backprojection reconstruction


Image registration

(Nuclear Medicine) See Registration

Image registration

(Radiotherapy) This is a multi-modality image processing procedure. In radiotherapy medical images are prepared in the required format for treatment planning. This usually involves manual or automatic fusion of CT-MR-PET images and delineation of target volume and tissue structures of interest.

Image retrieving

(General) Image retrieval is the process of recovering a particular image or set of images from a large data archive of images.

Within imaging departments, retrieval is usually limited to operating the software of the digital archive of medical images stored locally.

Image data are typically stored within a separate physical electronic data bank (PACS), with each image carrying associated details such as type of image, imager settings, and all relevant patient details.

Often such image archives are linked to the patient’s electronic record, so that when retrieving specific images in clinics, on the ward, or in the surgery, they can be identified and selected from the patient’s electronic record, and automatically retrieved and downloaded onto the physicians monitor.

In future, more sophisticated image retrieval systems should be able to assist in research applications. The use of ‘content-based image retrieval’ will allow images to be retrieved by their similarity with other images.

Image selected in vivo spectroscopy (ISIS)

(Magnetic Resonance) ISIS is one of the most common spatial localisation techniques used for single voxel spectroscopy (SVS). Because it involves acquisition of a free induction decay (FID) signal shortly after excitation, rather than a delayed echo, it is particularly suited to nuclear species with short T2 relaxation times, such as phosphorus (31P) nuclei.

The ISIS technique requires post-acquisition combination of FID signals acquired following each of eight dissimilar pulse sequences. One of these sequences is shown in Figure I.11. The
effect of this sequence is to invert magnetisation throughout three mutually orthogonal slices, so this magnetisation lies along the negative z-axis. Where two slices intersect, magnetisation experiences two inversion pulses and hence is returned to the positive z-axis. In the volume formed by the intersection of all three slices, magnetisation experiences three inversion pulses and the net effect is to place it along the negative z-axis. Signal acquired at the end of the sequence using the 90° pulse contains positive and negative components reflecting this pattern of spin preparation. The other seven sequences are obtained by omitting slice selection along each of the three axes in turn (three sequences), along each possible combination of two axes (three sequences) and along all three axes (one sequence). After each sequence, the spatial distribution of positive and negative z-magnetisation throughout the sample, and hence the composition of the FID signal, varies. Appropriate addition and subtraction of all eight signals should ideally lead to cancellation of signal outside the volume of interest (VOI) defined by the three intersecting slices, leaving only the VOI signal for Fourier transformation to yield the desired spectrum.

In practice, the process of adding and subtracting eight large signals to produce a relatively small localised signal is subject to a variety of sources of error. Variation in signal intensity due to movement or instrumental drift can result in contamination of the spectrum with signal from outside the VOI, and there can also be dynamic range and digitisation problems. If the read-out pulse is not precisely 90° over the whole sample and the sequence repetition time is too short for complete T1 recovery, further contamination arises due to a mechanism known as 'T2-smearing'. Under clinically realistic conditions, T2-smearing can result in up to 70% of the acquired spectrum originating from outside the VOI, although this can be reduced dramatically by judicious choice of VOI and acquisition parameters. Several variants of ISIS have been developed over the years, often with the aim of reducing contamination.

Related Articles: Point resolved spectroscopy (PRESS), STEAM, Magnetic resonance spectroscopy, Single voxel spectroscopy


Image sequences

(Diagnostic Radiology) Image sequences are used in various x-ray examinations to record quick dynamic changes (most often during x-ray angiographic examinations). One sequence (series) could have approximately two to eight images (exposures) per second. This requires powerful x-ray generator capable of producing short exposures with sufficient power. Often the sequence of exposures is produced by grid-controlled x-ray tube. The image sequence can be synchronised with the angiographic injector or with the ECG (in cardioangiography), this way specific phases of the examination (as arterial phase, capillary phase, venous phase) can be recorded. One examination can be programmed to have a number of image sequences separated by pauses. The image sequence (known also as serial exposures) can be recorded either with a spot film camera (in older systems) or with a digital detector (in newer systems as DSA, where the images per second can be more than 30/s).

Related Articles: Serial exposures, Digital subtraction angiography (DSA)

Image smoothing

(Nuclear Medicine) Image smoothing is a special case of filtering performed to reduce the impact of noise. Since noise is a high frequency phenomenon most of the image smoothing filters are low-pass filters that enhance low frequencies relative to higher frequencies. The prominent downside when using smoothing filters is the loss of spatial resolution.

Image storage

(Diagnostic Radiology) See Picture archiving and communication systems (PACS)

Image uniformity

(Magnetic Resonance) Image uniformity is a measure of the extent to which an MR image faithfully reproduces a perfectly uniform imaged object. It reflects the performance of the imaging system as a whole: hardware, imaging pulse sequence and reconstruction software, and can also be affected by features of the object such as magnetic susceptibility differences and dielectric properties. See Non-uniformity

Related Article: Non-uniformity

Image-guided brachytherapy

(Radiotherapy, Brachytherapy) In image-guided brachytherapy, IGBT, 3D images of the patient with the applicator(s) inserted are used to define tumour extent, target volumes, organs at risk, and the applicator(s) with source and stop (dwell) positions, as the basis for the treatment planning.

In conventional external beam radiotherapy set-up uncertainties, etc. are compensated for by margins added to the CTV, the clinical target volume, resulting in the planning target volume, the PTV, a purely geometrical concept.

The situation in brachytherapy is different since the PTV can often be regarded as being the same as the CTV because the inserted applicators are stable with respect to the CTV.

The imaging technique chosen for IGBT should in principle be able to show, in a single 3D image data set, the tumour and the target volume(s), organs at risk as well as the applicators and the source stop (dwell) positions. Imaging techniques must be adapted to suit the requirements of the specific brachytherapy techniques used and organ of interest, e.g. T2-weighted MR imaging for cervix cancer tumour definition and MR and CT compatible applicators must be used. It should also be noted that accurate determination of the dose distribution,
Image-guided radiotherapy

(faithful applicator reconstruction and representation of source stop (dwell) positions all depend on the selection of appropriate imaging techniques.

Generally speaking, for HDR brachytherapy with the applicators firmly held in place during the whole procedure, the movements of applicators relative to target and organs at risk should be minimal, i.e. the dose delivered should correspond to the dose planned.

Figure I.12 shows a simple example of an image-guided brachytherapy using CT. This is an intracavitary treatment of a residual maxillary cancer using a mould technique with three catheters. The patient had been treated previously with pre-operative external beam radiotherapy and surgery, and the residual tumour is being treated with brachytherapy. The mould had been fabricated specifically for this patient and tested (it was inserted in place in the maxillary cavity though the mouth, as the floor of the mouth had been removed at surgery), optimal catheter positioning was discussed by the treatment team (head-and-neck surgeon, oncologist, medical physicist), and channels for the catheters inserted into the mould. A preplan was also produced before the patient’s treatment to ensure that the geometry and overall treatment plan were correct.

For the first brachytherapy fraction, a 3D CT image with the applicator in place was used to define the target volume, the applicator, catheter and corresponding source stop (dwell) positions. A treatment plan was created and accepted, and the patient treated. Orthogonal radiographs were also taken and used to verify applicator position in relation to bony structures for the remaining three fractions.

**Abbreviations:** CT = Computed tomography, CTV = Clinical target volume, HDR = High dose rate, MR = Magnetic resonance and PTV = Planning target volume.

**Related Articles:** Volumetric prescribing – brachytherapy, Interstitial brachytherapy, Interactive implant technique

**Image-guided radiotherapy**

(Image-guided radiotherapy) Image-guided radiotherapy (IGRT) is a term used to describe several approaches to using imaging in radiation therapy. The first of these is the use of functional imaging, particularly SPECT/PET and MRI, to aid target volume definition. The second is the use of soft tissue image to aid verification at time of treatment for external beam radiotherapy. The third is the use of anatomical imaging to assist in the planning and source implantation for treatments such as brachytherapy.

**Image-Guided Planning:** Image-guided planning involves the use of functional positron emission tomography (PET), single photon emission computed tomography (SPECT) or magnetic resonance imaging (MRI) to help define the target volume for radiotherapy planning. Often the functional tracers used are markers for metabolism (as in 18F-FDG PET imaging). More recently markers for hypoxia (e.g. 60Cu-ATSM PET) and cell proliferation (e.g. FLT PET) have been developed and have started to be used. Often the data from these functional modalities are combined with anatomical imaging using CT to maximise the information used to define the target volume. Radiotherapy planning is done on CT data and all other imaging modalities have to be registered to the CT so as to give clinicians information to help them define the target volume.

![Figure I.12](See colour insert.) Image-guided brachytherapy: Images of a patient with a maxillary cancer, previously treated with pre-operative external beam radiotherapy and surgery, show a brachytherapy insertion and dose distribution to the residual tumour. A three channel mould applicator has been used with CT-based treatment planning (BrachyVision, Varian).
Using a variety of medical imaging modalities.

Clinical procedures: Ultrasound, Brachytherapy, Conebeam CT

Functional imaging, Treatment verification, Electronic portal imaging

Radiotherapy for verification. These broadly fall into two categories:

1. The use of internal fiducial markers as a surrogate for the tumour (imaged with kV or MV x-rays in portal imaging); and the imaging of soft tissue directly. The latter is often achieved using conebeam CT, or CT on rails, or using ultrasound – all in the treatment room.

2. Another approach to the fiducial marker method is to use implanted electromagnetic transponders whose positions are located using a receiver. Figure I.13 shows an example of a conebeam CT imaging system for IGRT verification.

Related Articles: Portal imaging, Set-up error, Stereotactic frame

Imaging

(General) Imaging is the process of producing visual representations of physical objects. Medical imaging is the process of producing images of internal body anatomy, conditions and functions using a variety of medical imaging modalities.

Imatron

(General) Imatron is a vendor name of electron beam CT (see the eponymous article).

Related Article: Electron beam CT

Immobilisation

(Radiotherapy) Immobilisation (or fixation) of the patient is an essential aspect of radiotherapy, enabling the accurate reproduction of the treatment position. This will reduce the dose to normal tissue, allowing the prescribed dose to be escalated for greater tumour control.

A simple example of immobilisation is the use of a rigid couch surface continually throughout imaging, planning and treatment. Routines such as voluntary bladder emptying can be an effective and practical way of minimising internal changes in organ placement. Positioning systems also play an important part, and use skin contour systems, body tattoos, or a reference image using non-invasive infra-red or video based camera systems. Immobilisation devices can either be patient specific devices such as thermoplastic shells, or standard devices that can be tailored to a particular patient by rotation and angling.

The immobilisation device must be chosen carefully at the beginning of treatment planning, as the choice must be comfortable and easy to use, as will be used continuously all throughout imaging, simulator and treatment. Evaluation studies of immobilisation devices allow determination of volume margins for planning, and action levels for verification processes such as portal imaging. Immobilisation is especially important with the rise of IMRT, where the conformal dose distribution creates steeper dose gradients, often close to OARs. Patient set up errors will therefore have a greater effect on clinical outcome.

One of the most sophisticated immobilisation devices commonly used are stereotactic frames which gives <1mm accuracy and is used for radical brain radiotherapy.

Related Articles: Portal imaging, Set-up error, Stereotactic frame

Implant

(Magnetic Resonance) Metallic passive implants may cause serious effects, which include torque, heating on the implants and artefacts on MR images. Before imaging patients with MR any surgical procedure that the patient underwent prior to the MR examination must be assessed and the presence and characteristics of the implants has to be determined. Ferromagnetic implants should be avoided. Implants that involve magnets such as magnetic sphincters, stoma plugs, dental implants, etc. can be demagnetised by the MR procedure. They should be removed prior to the examination. Active implants are generally considered an absolute contraindication for MRI due to the risk of severe patient damage or death. The screening procedure for patients is one of the most critical components of a program to permit the safety of all those preparing to undergo MR procedures or to enter the MR environment. It should be noted that having undergone a previous MR procedure without incident does not guarantee a safe subsequent MR examination. Various factors (e.g. static magnetic field strength of the MR system, orientation of the patient, orientation of a metallic implant or object) can substantially change the scenario and therefore, a comprehensive screening procedure must be conducted any time a patient prepares to undergo an MR procedure.

Related Article: Metallic implant

Implant dose distribution

(Radiotherapy, Brachytherapy) The dose distribution for an interstitial implant or intracavitary insertion should be based on true
position of applicators/sources in relation to the volumes of interest, i.e. tumour, target and organs at risk.

**Film Based Implant Dose Distributions:** When dosimetry systems are used for predictive treatment planning, the actual applicator and source positions must be determined and approved before the treatment is started. Radiographs are generally used to determine the position of sources and the markers used to identify organs at risk, if any, in 3D. At least two projection radiographs with known geometry are necessary to define a point in three dimensions and orthogonal radiographs are most commonly used for the geometric reconstructions. It is important to note, that soft tissues cannot be visualised on these films. Starting from a reconstruction of applicators or sources, the implant dose distribution can be calculated in 3D using a treatment planning system with a brachytherapy source model. All brachytherapy dose distributions are calculated in 3D, even if the source model itself is nominally characterised as one dimensional.

**Image-Guided Implant Dose Distributions:** In modern image-guided brachytherapy a 3D image data set is acquired with the applicators in treatment position. This allows simultaneous 3D definition of tumour, target, organs at risk, applicators and source positions. The 3D dose calculations make it possible to evaluate the implant dose distribution in terms of dose volume histograms of parameters of tumour and target volumes as well as organs at risk. Although brachytherapy source models still have a number of limitations, image-guided brachytherapy represents a great advance.

**Abbreviation:** 3D = Three dimensional.

**Related Articles:** Treatment planning systems – brachytherapy, Source models, Orthogonal films, Dose volume histograms – brachytherapy


**Impulse response function**

*(Nuclear Medicine)* The impulse response function (IRF) describes imaging systems degradation because of its inherent limitations. Another common word is the point spread function (PSF) or point response function (PRF). The idea behind determining the degradation is to image a source that is much smaller than the spatial resolution of the system. For SPECT systems and scintillation cameras, one often uses a sealed source such as Co-57. The full width at half maximum (FWHM) is determined from the image of the source and will provide information about the system spatial resolution. The modulation transfer function is calculated from the impulse response function (IRF) and may sometimes provide a better understanding of the degradation in image quality.

**In-111-labelled ibritumomab tiuxetan (Zevalin®)**

*(Nuclear Medicine)* See Y-90-labelled ibritumomab tiuxetan (Zevalin®)

**In air calibration factor**

*(Radiation Protection)* The ionisation chamber is recommended by international protocols for determination of the absorbed dose in photon and electron beams. In air calibration factor relates the dose in the chamber gas to the measured collected charge. The determination of this calibration factor for the parallel plate ionisation chamber is obtained from a comparison of the absorbed dose to water value measured with a cylindrical reference chamber, as recommended by international protocols IAEA TRS 381, IAEA TRS 398.

**Abbreviations:** IAEA = International Atomic Energy Agency and TRS = Technical reports series.

**Related Articles:** Dose, Radiation dosimetry


**In vivo body composition**

*(Radiotherapy)*

**Background:** The body comprises distinct and measurable body compartments. The status and rate of change of these compartments reflects the health of a person and the response of treatment for a specific disease. Whereas measurement of body weight (M) is a basic and useful parameter, it can be a misleading measure of response to treatment, which may increase oedema and fat while depleting protein.

The body compartments are as follows: total body protein (TBP), total body nitrogen (TBN), total body potassium (T BK), total body water (TBW), total body fat (TBF), total body calcium (TBCa), total body carbon (TBC) and fat free mass (FFM). FFM includes all body tissue excluding fat: i.e. skeletal muscle, visceral organs, bone and skin, and body water, as well as hair, blood, and lymph.

Compartments are related as follows:

\[
\text{TBP} = 6.25 \times \text{TBN}
\]

FFM is calculated from TBK on the basis that the potassium content of FFM is 2.26 g/kg in females and 2.52 g/kg in males, as potassium is not found in adipose tissue.

However, changes in TBK can reflect chemical changes in the body and blood. For this reason TBN is regarded as a superior measure of protein status in diseased subjects.

The mass of a subject is the sum of the defined compartments in the two and four body models:

\[
\text{M} = \text{FFM} + \text{TBF}
\]
\[
\text{M} = \text{TBP} + \text{TBF} + \text{TBW} + \text{TBCa}
\]

**Methods:** In vivo methods are used to obtain values of the body compartments. These might be done before and after therapy, to determine the effect of therapy. The measurement of normal values is required in order to compare patient values, according to defined indices. For example, the nitrogen index \(\text{NI} = \frac{\text{TBN}}{\text{TBN}_n}\), where \(p\) designates the patient and \(n\) the normal values for healthy subjects.

The importance of such indices relates to the impact on disease prognosis.

In vivo interrogation techniques are used. These can give a measure of the whole compartment, or in some cases the spatial distribution of the compartment. Analysis of blood and urine after ingestion of isotopically labelled molecules can also be used:

1. The major clinical application of in vivo body composition analysis rests with the dual energy x-ray (DEXA) technique for the determination of bone density, of particular relevance to the management of osteoporosis, especially in the aging population. There is a very low and insignificant radiation dose associated with this measurement (∼10 μSv).
2. Computed Tomography (CT) can be used to measure bone density but invokes much higher and significant radiation doses.
3. High energy neutron inelastic scattering in calcium can give the TBCa by measurement of the characteristic inelastic gamma ray, but at a high radiation dose.

4. Nitrogen is measured in a body protein monitor (BPM) using a CT252 or PuBe neutron source. The patient is moved over a collimated neutron beam. The fast neutrons emitted by these radioactive sources are moderated in tissue and captured by hydrogen and nitrogen nuclei in the patient. The 1.46 MeV ground state gamma ray from nitrogen capture is weak but can be measured by NaI detectors as I has higher energy than all background radiations. The hydrogen gamma ray is 2.2 MeV but the yield is intense and easily measured. The ratio of N/H is independent of body habitus.

5. Potassium-40 has a half-life of 1.3 × 10⁸ years, and is present with 0.0117% abundance in all potassium. The 1.46 MeV gamma ray is readily detected by an array of NaI detectors. There is no additional radiation dose, apart from the naturally occurring potassium. A low background, shielded room is required for accurate measurements with old steel (forged prior to the A-bomb testing program).

**Abbreviations:**
- BMI = Body mass index
- BPM = Body protein monitor
- DXA = Dual energy x-ray absorptiometry
- FFM = Fat free mass
- LBM = Lean body mass
- NI = Nitrogen index
- TBC = Total body carbon
- TBCa = Total body calcium
- TBF = Total body fat
- TBK = Total body potassium
- TBN = Total body nitrogen
- TBP = Total body protein
- TBW = Total body water

**Related Articles:**
- Total body protein
- Total body nitrogen
- Total body potassium
- Total body water
- Total body fat

**In vivo dosimetry**

*(Radiotherapy)* In vivo dosimetry is the verification method for checking dose delivery directly during the treatment. It is performed to detect errors in individual patients, to detect errors in core procedures, to evaluate the quality of specific treatment techniques or to evaluate the dose in situations in which the dose calculation is inaccurate or not possible (e.g. non-standard SSD or using bolus).

In vivo dose measurements can be divided into entrance dose measurements, exit dose measurements and intracavitary dose measurements.

**Entrance dose measurements** serve to check the output and performance of the treatment apparatus as well as the accuracy of patient set-up. If entrance dose measurements alone are applied, the entrance dose has to be converted to the corresponding target dose using patient and treatment set-up information.

**Exit dose measurements** serve, in addition, to check the dose calculation algorithm and to determine the influence of shape, size and density variations of the body of the patient on the dose calculation procedure. In the real patient there is in most cases a considerable loss of backscatter, while the TPS calculations are valid for semi-infinite patients implying complete backscatter at the exit surface. A correction is then necessary.

A combination of entrance and exit dose measurements is a more accurate method of obtaining the target dose. Various methods are available to obtain the midline dose from entrance and exit dose values. These methods give generally good results for homogeneous situations but in the presence of inhomogeneities considerable deviations can occur.

In vivo dose measurements not only serve to check the dose delivery to the target volume but are also applied to assess the dose to organs at risk (e.g. the eye lens, gonads and lungs during TBI).

Portal imaging systems can also be used for in vivo dosimetry. Portal images can be transformed to ‘dose images’, which can then be correlated with exit dose values. Various groups are currently studying the usefulness of films or EPIDs for in vivo dosimetry. Forward approach and backward approach can be used. The relationship between the exit dose and the transmission dose at the position of the portal imaging detector is not simple and depends on many factors, such as the skin to detector distance, field size, patient thickness and photon beam energy.

Since a relatively large number of images can be made during one treatment fraction, EPIDs can be used to measure the influence of organ and patient motion on the dose distribution during one treatment session. Portal dose measurements are extremely useful in detecting differences between actual patient data as encountered during treatment and those applied during treatment planning. EPIDs are likely to become very useful for dosimetric quality assurance of intensity-modulated beams.

For in vivo dosimetry, the most commonly used detectors are silicon diodes, MOSFET dosimeters and TLDs.

**Abbreviations:**
- EPID = Electronic portal imaging device
- MOSFET = Metal oxide semiconductor field-effect transistor
- SSD = Source surface distance
- TBI = Total body irradiation
- TLD = Thermoluminescent dosimeter
- TPS = Treatment Planning System

**Related Articles:**
- Entrance dose
- Exit dose
- Dose verification
- Diode detectors, TLD

**Further Readings:**

**Inch**

*(General)* The inch is a small Imperial and US unit of length, being equivalent to exactly 25.4mm, and one twelfth of the larger unit of length, the ‘foot’.

The abbreviations ‘ft’ and ‘in’ are commonly used, as in 3 ft 6 in., and also in shortened form as in 3½.

**Incidence angle**

*(Radiotherapy)* See *Oblique incidence*

**Related Articles:**
- Oblique incidence
- Electron oblique incidence
- Obliquity
- Obliquity effect

**Incident dose**

*(General)* This is an ambiguous term used to describe the absorbed dose (usually in air) or air kerma or exposure incident on an absorber. See related terms.

**Related Articles:**
- Absorbed dose
- Air kerma
- Exposure

**Incident energy fluence**

*(Radiation Protection)* This is not a recognised radiation unit. It is used to describe the energy fluence incident on an absorber.

**Related Article:**
- Energy fluence

**Incoherent scattering**

*(Radiation Protection)* Interaction between an incident photon and a loosely bound atomic electron in which only some of the energy of the photon is transferred to the electron. The photon is scattered into a new direction, and the electron, referred to as Compton or recoil electron, is ejected from the atom with kinetic energy equal to the loss of energy of the photon.

Also known as inelastic scattering, Compton interaction, or Compton scattering.

For more information, see *Compton effect*.

**Related Article:**
- Compton effect
**Indirect detection**
(Radiation Protection) Photons and neutrons are indirectly ionising radiation, that is they react with the medium (the matter the radiation is passing through) producing charged particles that are ionising.

For photons, the photon(s) beam passing through matter can interact with orbital electrons (photoelectric effect), producing photoelectrons, or scatter, losing some of its energy to an orbital electron which is ejected from the atom (Compton effect). If the photon energy exceeds 1.022 MeV (i.e. the sum of rest mass of electron and positron) the photon may interact with the atomic nuclei and an electron–positron pair created.

Neutrons are detected using nuclear reactions such as $n, (n,\alpha), (n, p), (n, fission)$ or $n, (n, \gamma)$, in which case a radioactive isotope is produced.

Also, the indirect detection of ionising radiation is considered as based on the changes induced by radiation in the detector that are not observable immediately (directly). The intensity of radiation measured for estimating, e.g. radioactivity of the sample or dose, may be read after the exposure and not during exposure as it is possible with the use of ion chamber, scintillation or semiconductor counters.

The monitoring of the radiation exposures for safety purposes of the staff is often realised with use of indirect (passive) detectors, such as film badges where the optical density of the film changes when irradiated. After developing the optical density of the film it is measured, which with correct calibration, can be related to the radiation dose. The thermoluminescent detector (TLD) stores the absorbed energy of ionising radiation and when heated visible optical radiation is emitted, intensity of which is proportional to the radiation dose and can be measured, for example using a photomultiplier.

The alanine radiation detector (aminopropionic acid) is a dosimeter in which ionising radiation produces free radicals and concentration of the free radical is measured using the EPR (electron paramagnetic resonance) technique. The reading does not destroy the record, as in case of TLD.

Neutron detection is based on proton recoil or alpha particles recorded, for example by track-etch detectors which are special foils in which ionising radiation produces free radicals and concentration of the free radical is measured using the EPR (electron paramagnetic resonance) technique. The reading does not destroy the record, as in case of TLD.

**Abbreviations:** EPR = Electron paramagnetic resonance and TLD = Thermoluminescent detector (Dosimeter).

**Related Articles:** Absorbed radiation, Alpha particles, Compton effect, Dose, Dosimeter, Film badge, Gamma radiation, Ionisation chamber, Neutrons, Pair production, Photoelectric effect, Scintillation detector, Personal dosimeter


**Indirect digital radiography**
(Diagnostic Radiology) Indirect digital radiography refers to digital systems with flat panel, where x-rays are first converted to light, which is then detected by photodiodes or CCD (see the article *Flat panel detector*).

Historically x-ray film radiography has been considered a direct method, while image intensifier systems has been considered indirect (the signal passes several transformations). Later the digital systems have passed through similar divisions – direct digital radiography (x-rays transferred to charge by amorphous selenium) and indirect digital radiography (using phosphor and photodiode or CCD).

**Related Article:** Flat panel detector

**Indium-111** [111In]
(Nuclear Medicine)
Element: indium (group III-A)
Isotopes: 51 < N < 84
Atomic number (Z): 49
Neutron number (N): 62
Symbol: 111I

Production: cyclotron, e.g. $^{111}\text{Cd}(p,n)^{111}\text{In} \rightarrow ^{111}\text{Cd}$ or $^{112}\text{Cd}(p,2n)^{111}\text{In} \rightarrow ^{111}\text{Cd}$

Daughter: 111Cd
Half-life: 2.8 days
Decay mode: EC – decay
Radiation: gamma, internal conversion electrons, Auger electrons, characteristic x-ray photons
Gamma energy: 171.28 keV (90%) 245.42 keV (94%)
Dose rate from 1 MBq: 0.0842 µSv/h at 1 m; 842 µSv/h at 1 cm
Absorption (HVL): 0.8 mm lead
Biological half-life: 70 days
Critical organ: red bone marrow, liver, spleen, testes and lymph nodes
ALI, int (50 mSv): 200 MBq
Absorbed dose (chloride): 0.88 mGy/MBq kidneys, 0.60 mGy/MBq liver, red bone marrow.
Effective dose: 0.20 mSv/MBq (111InCl3); 0.40 mSv/MBq (111In-labelled leukocytes, platelets)

### Clinical Applications
Indium-111 has been used in nuclear medicine since the late 1970s, and has become the standard radionuclide for labelling of different biomolecules, such as blood cells (leukocytes and platelets) pentetreotide, and monoclonal antibodies using different linker molecules such as bifunctional chelates.

Today, the most known 111In-labelled radiopharmaceuticals are 111In-pentetreotide (111In-Octreoscan”, Mallinckrodt Medical) for localisation of primary and metastatic neuroendocrine tumour cells expressing somatostatin receptors and 111In-lbritnomam tiuxetan (111In Zevalin”, Schering AG) for imaging of relapsed or refractory follicular non-Hodgkin’s lymphoma, prior to radioclonide therapy with 90Y-Zevalin.

The chemistry of indium resembles that of iron and gallium, and a close similarity between indic and ferric ions has been established. Elimination of indium activity from organs and tissues is extended because of trapping by the plasma protein transferrin. Ionic 111In“ injected at pH <3.5, 111In-chloride or 111In-citrate, are rapidly captured by transferrin in the vascular system. Thus, 111In activity is distributed heterogeneously in the body and may locally contribute
significantly to the absorbed dose, since even though indium is a foreign element in our body, many different tissues show affinity for indium (transferrin receptors), especially rapid proliferating tissues.

The activity of $N^{13}$ and $O^{15}$ per unit volume of air produced in the treatment room by a $20\text{cm} \times 20\text{cm}$ beam is given by

$$ C = \frac{(D/16.7)(P)(1-e^{-\lambda T_r})}{V(\lambda + Q/V)}(e^{-\lambda T_r}) $$

where

- $C$ is the concentration (Bq/m$^3$)
- $D$ is the dose rate at maximum build up depth (mGy/s)
- $P$ is the production rate of $N^{13}$ or $O^{15}$ of 16.7 mGy/s (Bq/kg of oxygen or nitrogen)
- $Q$ is the room ventilation rate (m$^3$/s)
- $V$ is the volume of the room ($m^3$)
- $\lambda$ is the decay constant for $N^{13}$ and $O^{15}$ (s$^{-1}$)
- $T_r$ is the irradiation time (s)
- $T_{ir}$ is the time after end of irradiation (s)


Inelastic scattering (Radiation Protection) Interaction between an incident photon and a loosely bound atomic electron in which only some of the energy of the photon is transferred to the electron. The photon is scattered into a new direction, and the electron, referred to as Compton or recoil electron, is ejected from the atom with kinetic energy equal to the loss of energy of the photon.

Also known as inelastic scattering, compton interaction, or Compton scattering.

For more information, see Compton effect.

Related Article: Compton effect

Inflow effect (Magnetic Resonance) In sequences using two or more spatially selective RF pulses, such as the SE and IR sequences, the signal is sensitive to the transport of spins into and out of the excited slice. If the flow is coherent over at least a number of voxels (macroscopic flow) and has components perpendicular to the imaging slice, the signal will decrease due to the outflow of spins during the time between RF pulses (wash-out effect). On the other hand, if the repetition time is not too long, the inflow of ‘fresh’, non-saturated spins during the repetition time, will increase the magnetisation compared with the static, saturated, spins and thus give an increase in signal – an effect called the inflow effect or the wash-in effect (in early literature also known as ‘paradoxical enhancement’).

These competing mechanisms make the modulus signal behaviour versus flow velocity biphasic and possibly difficult to interpret, although, with a rough knowledge of the velocities involved, the experimental parameters could be adjusted so that one mechanism is dominant. If, however, only one RF pulse is used per repetition, as in gradient echo (GRE) sequences, the signal decrease due to washout will not occur and an enhanced signal will result from the inflow effect. Ideally, for plug flow perpendicular to the imaging slice and

### Inflow effect

**Related Articles:** Gallium-67, Ytrium-90 ibritumomab tiuxetan (Zevalin)

perfect slice shape, the signal increase will be linear with velocity up to the velocity when all spins are exchanged in the slice during TR (see the following figure).

![Diagram of signal increase with velocity](image)

**Related Article:** Flow void

**Infrared radiation**
*(Radiation Protection)* Infrared radiation is otherwise known as heat radiation and is the part of the electromagnetic spectrum between visible light and microwaves/radiowaves. The IR spectrum ranges from a wavelength of 0.7 μm at the red end of the visible spectrum, to around 1 mm, and is therefore in the non-ionising region of the spectrum.

**Related Articles:** Electromagnetic radiation, Non-ionising radiation

**Inherent contrast**
*(Magnetic Resonance)* Inherent contrast refers to the contrast properties present in an image without the use of an exogenous contrast agent.

Inherent contrast in an MR image may depend on a wide range of variables, including the concentration of water protons and their physicochemical environment (T₁ and T₂), bulk flow and perfusion of blood, and the rate of water molecule diffusion. These variables, in turn, may reflect a variety of physiological parameters, such as water content, temperature, brain activation and tumour angiogenesis.

The extent to which each of these variables influences the appearance of an image is controlled through pulse sequence design, either by simple changes to sequence parameters (e.g. shortening the repetition time, Tₑ, to increase T₁ weighting) or by adding sequence elements (e.g. Stejskal–Tanner gradients for diffusion sensitisation).

Thus it is possible to alter the inherent contrast and associated information content of an MR image to an essentially limitless extent in order to answer different clinical questions. This capability makes MRI far more flexible than most imaging modalities.

**Related Articles:** Contrast agent, Unenhanced image

**Inherent filtration**
*(Diagnostic Radiology)* The total filtration of an x-ray beam aims to reduce the unnecessary low energy x-ray photons. It is produced by a combination of two filter components – inherent and added. The inherent filtration is due to existing components of the x-ray tube and housing through which the x-ray beam passes (tube glass, oil in tube housing, beam locator mirror, etc.). After removing the added filtration, the inherent filtration can be measured by means of HVL.

**Related Article:** Total filtration

**Inhomogeneity**
*(General)* Inhomogeneity refers to field, image, etc. which is not uniform (homogeneous). See specific articles related, for example to field homogeneity in MRI; image uniformity in diagnostic radiology or nuclear medicine; dose distribution in radiotherapy, etc.

**Inhomogeneity correction factor**
*(Radiotherapy)* For accurate dose calculations in a heterogeneous volume such as a patient, corrections must be made for the presence of non-water-equivalent tissue, or inhomogeneities.

One approach to doing this starts with the dose distribution for a homogeneous water equivalent density and then applies correction factors to change this distribution to account for the different tissue densities. In this approach, the dose in the heterogeneous case is given by the following equation:

\[
D_{\text{het}} = D_{\text{water}} \times \text{ICF}
\]

Equation I.1 shows how to correct a homogeneous dose distribution for the presence of an inhomogeneity using inhomogeneity correction factors (ICF).

where

- \(D_{\text{het}}\) is the dose distribution in heterogeneous tissue
- \(D_{\text{water}}\) is the dose distribution in a water volume
- ICF is the inhomogeneity correction factor

There are a number of methods for calculating ICF including approaches such as effective attenuation coefficient, power law (Batho), ratio of TAR and equivalent TAR. These are summarised in a number of textbooks and reports including APPM (2004).

**Abbreviations:** ICF = Inhomogeneity correction factor and TAR = Tissue air ratio.

**Related Article:** Heterogeneity

**Further Reading:** AAPM. 2004. Tissue inhomogeneity corrections for megavoltage photon beams, AAPM Report number 85, Medical Physics Publishing, Madison, WI.

**Inorganic phosphate**
*(Magnetic Resonance)* Inorganic phosphate features in in vivo phosphorus (³¹P) NMR spectra. The term encompasses four distinct species – the phosphate ion \(\text{PO}_{4}^{3-}\) (Figure I.14), the hydrogen phosphate ion \(\text{HPO}_{4}^{2-}\), the dihydrogen phosphate ion \(\text{H}_{2}\text{PO}_{4}^{-}\), and phosphoric acid \(\text{H}_{3}\text{PO}_{4}\).

These species exist in pH-dependent equilibrium, and under physiological conditions inorganic phosphate occurs as \(\text{HPO}_{4}^{2-}\) and \(\text{H}_{2}\text{PO}_{4}^{-}\). These ions have different chemical shifts, and so the position of the inorganic phosphate peak in a ³¹P spectrum depends on the relative proportions of the two species and hence on pH. Thus the chemical shift difference between inorganic phosphate

![Molecular structure of the phosphate ion](image)

**FIGURE I.14** Molecular structure of the phosphate ion.
and phosphocreatine is frequently used to determine intracellular pH according to the following formula:

$$\text{pH} = \text{pK}_A + \left( \frac{\delta - \delta_A}{\delta_A - \delta} \right)$$

Here $\delta$ is the chemical shift of the $P_i$ peak relative to $PCr$, $\delta_i$ is the chemical shift of hydrogen phosphate (approximately 5.70 ppm), $\delta_A$ that of dihydrogen phosphate (approximately 3.23 ppm), and $\text{pK}_A = 6.77$ (Figure I.15).

Inorganic phosphate has a role in the body’s energy metabolism as a by-product of the dephosphorylation of ATP. Thus its concentration, and the size of the resonance peak, increases in muscle during exercise and in the brain during hypoxia/ischaemia.

**Inorganic scintillators**

(Nuclear Medicine) An inorganic scintillator is one out of two groups of scintillators (inorganic and organic). The purpose of scintillators is to absorb high energy radiation and in the process produce scintillation light that can be measured using a photomultiplier tube. This detection setting is the most common in modern emission imaging systems. Inorganic scintillation materials are solid crystals and the basic condition for scintillation comes from characteristics in their lattice structure, individual atoms or molecules do not scintillate. In organic materials on the other hand the scintillation is a molecular property rather than an effect of crystal structure.

The scintillation property of an inorganic scintillating material depends on the energy states determined by the lattice structure. The electrons in such materials are only allowed in certain discrete bands, the upper two bands are called valence band (the lower of the two) and conduction band (upper). In a pure crystal these two bands are separated by a gap of ‘forbidden’ energies. The emitting of scintillation light is part of a two step process; (1) incident radiation excites electrons in the lower valence band up to the conduction band, (2) electrons release energy when they are de-excited, thus sending out a scintillation photon. Some of these transitions can be radiationless, i.e. not producing any photons. Radiationless transitions are referred to as quenching. Quenching leads to a non-linear response between deposited energy and the number of photons produced, although the problem is minimal relative to quenching in organic scintillators.

A desired feature of a crystal is a high light yield, namely a high number of photons per energy deposit. A crystal with high light yield provides a better signal, hence better signal to noise ratio.

In most pure crystals the energy gap between the two bands, and therefore the energy of the emitted photon, is too wide to produce photons in the visible part of the spectrum. When using photomultiplier tubes to strengthen the signal the ultimate scintillation light is around 400 nm (blue light). A common approach used to create photons in the visible part of the spectrum is to add impurities to the lattice. These impurities create ‘islands’ of available energy states in the forbidden gap. The electrons will de-excite via the impurity induced energy states and emit photons with lower energies.

A desired crystal feature is that the half-life of each excited energy state is short. A long decay time will lead to dead time losses (see separate article) because the chance of two signals overlapping increases. Typical decay times are 50–500 ns.

The use of impurities also introduces an unwanted effect called phosphorescence or afterglow. Some electrons are ‘caught’ in energy states where further de-excitation is forbidden. The only way to de-excite to the ground state is to first excite to a higherlying energy state. The photons produced in such cases are delayed in time, which has a degenerative effect on the signal.

Crystals with impurities have no extensive self-absorption of the emitted radiation as is the case with pure crystals. The impurities are spread out over the lattice and work as emitter and absorber centres, so the probability of re-absorption is lower than in a pure crystal where all atoms can absorb the scintillation light.

**Related Articles:** Nal (TI) detector crystal, Scintillators, Phosphorescence, Light yield in scintillation detectors, Bismuth germanate (BGO)


**Input circuit**

(General) The part of a device at which the (input) signal is applied or where either a measuring sensor/transducer or a device under test is connected.

The input circuit of electronic (signal measuring or sensing) devices is characterised by the input impedance and input signal range. If the device under test is a passive device (not generating any kind of power), the input circuit contains a power source enabling measurement of electrical characteristics of the device under test.

**Input curves on anode-cooling charts**

(Diagnostic Radiology) See Anode-cooling curve

**Input impedance**

(General) See Input circuit

**Input screen in fluoroscopy**

(Diagnostic Radiology) See Image intensifier

**Insulation resistance**

(General) The resistance between two conducting parts (electrodes) that are under normal conditions are insulated from each other. Insulation resistance is of the order of megaohms [MΩ]. Proper insulation resistance is important for patient and staff safety in all medical devices, but especially in those which operate with high voltage (x-ray, CT, US).

**Integral dose**

(Radiotherapy) One way of comparing dose distributions for different quality beams is to calculate the integral dose for a given tumour dose.
The integral dose is a measure of the total energy absorbed in the treated volume and it is desirable to keep it as low as reasonably possible. Except the case of large volume irradiation integral dose is not a limiting factor in radiotherapy, mainly in the modern techniques. If a mass of tissue receives a uniform dose, then the integral dose is simply the product of mass and dose. However, in practice, the absorbed dose in the tissue is non-uniform so a mathematical formula is required to calculate it.

For a single beam of photons an approximate method accurate enough was introduced by Mayneord who formulated the following approximate expression:

\[ \Sigma = 1.44D_0Ad_{1/2} \left( \frac{d}{d_{1/2}} \right)^\frac{1}{2} \left( 1 + \frac{2.88d_{1/2}}{SSD} \right) \]

where
- \( \Sigma \) is the integral dose
- \( D_0 \) is the delivered dose
- \( A \) is the field area
- \( d \) is the total thickness of the patient in the path of the beam
- \( d_{1/2} \) is the depth of 50% isodose
- SSD is the source-surface distance and the expression inside the brackets takes into account the geometric divergence of the beam

The unit for integral dose is kilogram gray or simply joule. The integral dose depends on the field arrangement and the beam quality. In Figure I.16 the integral dose as a function of the beam quality for a tumor dose of 10 Gy at a depth of 12.5 cm in a 25 cm thick patient treated with parallel opposed beams, field size 10 cm \( \times \) 10 cm at an SSD = 100 cm.

The application for integrated backscatter has mainly been in cardiac application, where it has gained some success. Due to the structure and movement of the heart, the backscattered signal from heart tissue will vary over the cardiac cycle. Where the sound beam is parallel to the muscle fibres of the heart, little signal is received, whereas when the beam is perpendicular to the fibres, the signal power is increased. The opposite occurs to the angular dependence of frequency-averaged attenuation. The timing of these variations, as well as the relative change of integrated backscatter and attenuation, relate to various disorders like infarcts, ischaemia and myopathy.

**Integrated parallel acquisition technique (iPAT)**
(Magnetic Resonance) See iPAT (integrated parallel acquisition technique)

**Integrating dosimeter**
(Radiation Protection) The integrating dosimeter measures the dose of ionising radiation during a known period of time. Integrating dosimeters play a very important role in personnel monitoring as well as in radiation protection of the patient.

There are active and passive integrating dosimeters. Typical active integrating dosimeters are gas-filled detectors, for example small size ionisation chambers called ‘pocket chambers’ (Figures I.17 and I.18).

The passive integrating dosimeters are of different kinds: photographic films, solid state nuclear track detectors, thermoluminescent dosimeters, activation foils. See related articles for examples.
Related Articles: Dosimeter, Film badge, Finger ring dosimeter, Geiger–Müller (GM) counters, Ionisation chamber, Thermoluminescent dosimeter (TLD)


Integrating the healthcare enterprise (IHE)
(Diagnostic Radiology) IHE (integrated healthcare enterprise) is an initiative by healthcare professionals and the industry to improve the way computer systems in healthcare share information.

IHE promotes the coordinated use of established standards such as DICOM and HL7 to address specific clinical need in support of optimal patient care. Systems developed in accordance with IHE communicate with one another better, are easier to implement and enable care providers to use information more effectively (Figure I.19).

IHE enhances the quality of patient care, resulting in the following benefits:

- Safety through the reduction of medical errors
- Savings through lower implementation costs and more efficient workflow
- Satisfaction through better informed medical decisions and faster results for both patient and physician

Hyperlinks: www.ihe.net

Intensification factor (IF)
(Diagnostic Radiology) Intensification factor is a characteristic of a specific intensifying screen and film combination. It is the ratio of exposure directly to a film compared to the exposure to an intensifying screen-film combination required to produce the same optical density. IF has very little meaning or application in modern radiography because direct film exposure is not a useful reference.

Intensifying screen
(Diagnostic Radiology) Intensifying screens (or intensifying phosphors) are sheets of fluorescent materials placed in contact with the x-ray film in radiographic cassettes to perform two functions. It absorbs the x-radiation and converts a fraction of the absorbed energy to light. The light exposes the film. The advantage is that an image can be created with much less exposure than if the film were exposed directly by the x-radiation. At least two factors contribute to this effect. The intensifying screen is thicker and a more efficient x-ray absorber than the film emulsion and the increase in photon concentration produced by the x-ray to light conversion is more effective in producing film exposure.

Until the 1970s calcium tungstate was the typical fluorescent material in radiographic intensifying screens. It has been replaced by rare earth fluorescent materials containing lanthanum or gadolinium. Contemporary intensifying screens use yttrium. Screen-film systems are now quickly replaced by computed radiography systems.

Related Article: Screen film

Intensifying screen(s), rare earth
(Diagnostic Radiology) See Rare earth screen

Intensity
(Ultrasound) Intensity is defined as the energy flux crossing a surface per second, i.e. it is the power per unit area. It has units of J/m²/s or W/m².

Its value at any point is proportional to the pressure amplitude squared of the sound wave:

\[ i = 2 \, \rho \, c^2 / p_c = 2 \, p^2 / Z \] assuming plane waves

where
\[ \rho \] is density of medium
\[ c \] is speed of sound
\[ Z \] is the acoustic impedance

For a diagnostic ultrasound beam the acoustic intensity will vary along the length of the beam as the beam cross-section changes shape due to focusing, and energy is attenuated due to absorption...
and scattering. Intensity is therefore usually greatest nearer to the transducer and at transmit focal points, where all the energy is passing through a small cross-sectional area. Intensities are also usually greatest for non-scanned modes such as m-mode and pulsed wave Doppler, as the beam is then stationary within the tissue, and when pulse-average intensities are higher such as pulsed wave Doppler and colour Doppler where longer pulses are used.

Heating of tissue is proportional to the intensity making intensity an important parameter to know when considering the safety of ultrasound and the likelihood of thermal damage.

In order to accurately measure intensity it is necessary to make measurements over all frequencies that may be present, including the high harmonic frequencies that may exist due to non-linear propagation within the beam.

It is often useful to further define how intensity is measured by considering spatial peak or average intensities and instantaneous or time average intensities.

**Related Articles:** Pulse average intensity, Spatial peak intensity, Spatial average intensity, Time average intensity, Temporal peak intensity

### Intensity-modulated radiotherapy

(Radiotherapy) Intensity-modulated radiotherapy (IMRT) is a set of techniques of varying the intensity across the radiation field to deliver complex 3D dose distributions in external beam radiotherapy. It is found to be particularly useful in creating dose distributions with concavities, which are needed when the target is in close proximity to a dose limiting critical organ.

Conventional radiotherapy uses a set of beams with fixed intensity profiles delivered in a cross fire effect to achieve a high dose to the target with maximal sparing of adjacent tissues. In IMRT the intensity profile of these beams is shaped such that the combined effect achieves the desired dose distribution. The added complexity generated by the extra degrees of freedom achievable with beam profile shaping means that the conventional manual, forward or interactive planning approach is generally not the best manner to plan the treatment and an automatic approach called inverse planning or inverse optimisation is used. This is a mathematical method of generating the treatment plan, in which a prescription is specified by the treatment planner in terms of doses to tumour and dose limits to critical structures. The computer then plans the beam distributions that produce a dose distribution that best matches the prescription.

Delivery of IMRT is most often achieved on a standard linac using a multileaf collimator, either as a set of static fields of differing shape from each beam direction or by scanning the leaves across the field during irradiation in the dynamic MLC technique. Other delivery systems for IMRT exist including intensity-modulated arc therapy, tomotherapy and robotic radiotherapy.

**Abbreviations:** IMRT = Intensity-modulated radiotherapy.

**Related Articles:** Conformal radiotherapy, Multileaf collimator, Inverse planning, Forward planning, Interactive planning, Tomotherapy, Robotic linacs, Gamma knife


### Intensity reflection coefficient

(Ultrasound) See Reflection coefficient

### Intensity transmission coefficient

(Ultrasound) See Transmission coefficient

### Intensity weighted mean

(Ultrasound) In many clinical applications, it is useful to be able to determine the mean velocity in a vessel. In combination with a measurement of cross-sectional area, mean velocity will give the volume flow through a vessel.

In ideal circumstances, if there is uniform insonation of the vessel, the mean velocity is obtained from the *intensity weighted mean* of the Doppler spectrum, defined as

$$f(t) = \frac{\int P(f) f df}{\int P(f) df}$$

where $P(f)$ is the Doppler power spectrum.

The intensity weighted mean can be obtained by analogue means or, in modern systems, by digital analysis of the spectrum analyser.

If the sample volume does not encompass the entire vessel or if there is non-uniform insonation then the calculated intensity weighted mean will not accurately represent the mean velocity. Non-axial flow can also lead to errors. The time averaged mean velocity (TAMV) (Figure I.20) shows the mean of this value over five cardiac cycles.

### Interaction

(Radiation Protection) A reaction between radiation and matter. Interactions may be broadly divided into particle-particle reactions, atomic interactions (where the radiation interacts with atom-bound electrons) and sub-atomic interactions (where the radiation interacts with the atom’s nucleus).

### Interactive implant technique

(Radiotherapy, Brachytherapy)

**Source Handling and Loading:** Brachytherapy sources must be handled and loaded into the applicators for treatment, and many methods have been used over the time. These methods have been developed primarily to reduce the dose to the personnel, but also to improve the quality of the treatment itself.

**Interactive Implant Technique:** Example: Ultrasound-guided interstitial interactive permanent implant of Iodine-125 seeds, an intra-operative procedure – prostate cancer.
Permanent prostate implants are often performed with manual loading and manual afterloading techniques, as low energy sources are used. During these procedures, dose to staff from the sources themselves is very low (Figures 1.21 through 1.23). If fluoroscopy is used during the implant procedure, the dose to staff comes mainly from the fluoroscopy.

The patient is sleeping during the whole procedure (requires anaesthesia equipment), and the procedure is performed in an operating theatre. The interactive implant procedure consists of:

1. Patient positioning (suitable treatment table, including leg supports)
2. Requirement: identical patient position during the whole procedure!
3. Image collection ultrasound (US) unit with bi-plane rectal probe, stepper for the rectal probe
4. Definition of target and organs-at-risk (dedicated treatment planning system, TPS)
5. Treatment planning – and plan acceptance
6. Verification of source strength (dosimetry equipment – electrometer and well-type chamber, insert suitable for the seeds used)
7. Preparation of needles and seeds/strands (seed handling equipment, implantation needles)
8. Interactive implantation (ultrasound unit with bi-plane rectal probe, stepper for the rectal probe, template to guide the needles, fluoroscopy unit, dedicated interactive treatment planning system)
9. Source positions are adjusted in the TPS according to US and fluoroscopy, the dose distribution is interactively updated
   a. 50–100 sources used
   b. Inter-source effects; not included in the dose calculations!
10. Book-keeping of sources – total number accounted for
    a. Implanted sources counted using fluoroscopy

**Related Articles:** Brachytherapy, Source loading in brachytherapy, Temporary implant, Permanent implant, Interstitial brachytherapy

**Interactive planning**

(Radiotherapy) Radiotherapy treatment planning involves choosing the beam directions, field shapes and intensities for the patient’s beam delivery. This is usually done using a CT scan of the patient and a model of dose deposition in the patient by the beam. A common approach to this is the interactive, or trial and error approach, in which beam parameters are tried by the planner and the resulting dose distribution evaluated in terms of target coverage and dose to non-target tissues. Often dose volume histograms are used as part of the evaluation of the treatment plan. This is also sometimes referred to as forward planning. The alternative to this is inverse radiotherapy planning.

**Abbreviation:** CT = Computed tomography.

**Related Articles:** Treatment planning, Inverse radiotherapy planning, Dose volume histogram
Interface
(Ultrasound) In ultrasound contexts the word interface is used when discussing borders between two materials with different acoustic impedances, for example the transducer-skin interface and blood-myocardium interface.

Related Articles: Reflection coefficient, Acoustic impedance, Reflection

Interference
(Ultrasound) Two waves that travel together can, dependent on their respective phase, be observed as a wave that is the sum of the two waves’ individual amplitudes (constructive interference), or add up to no apparent wave motion at all if the amplitudes are equal and the phases are opposite (destructive interference). In Figure 1.24 two point sources emit continuous waves, and in certain directions the waves appear to be in phase, whereas in others, to be out of phase. As can be deduced from Figure 1.24, this depends on the difference in distance from the observation point to the respective sources.

Interlacing monitors
(Diagnostic Radiology) See Acquisition modes for digital image

Interleaved
(Magnetic Resonance) In MRI the data collection can be interleaved in four different ways.

Interleaved k-space Coverage: The k-space can be acquired on one or more shots. When using a multi-shot technique the lines are often sampled interleaved. The first acquisition collects a series of lines in k-space and the second collects lines in between the first lines (see Figure I.25). Motion between shots has to be avoided.

Interleaved Slice Acquisition: When different slices of a scan overlap with each other a Crosstalk artefact can appear. This is caused by a slice profile that is not ideal due to constrain of the measurement technology. To avoid crosstalk the slice gap can be increased or the slice order may be interleaved. With interleaved slices the slices are not collected one by one in a row but in an
Interlock; Interlocking device

(Radiotherapy) The design of radiotherapy treatment machines and rooms includes many interlock safety features, in order to ensure that staff are not exposed to high doses, and as a fall-back to limit the consequences of any malfunction of the machine. Interlocks are examples of active engineered controls, which monitor a changing situation and can trigger a safety action, either electronically, or mechanically.

Common examples of interlocks are as follows:

• The ‘last man out’ button and door interlock. The machine will not irradiate unless the ‘last man out’ button has been pressed and the door is closed. The position of the ‘last man out’ button is such that it gives a clear view for the operator to ensure that no other person remains in the room, other than the patient. If the door is opened during irradiation, then the beam will immediately be turned off.

• Interlocks are used extensively with the computer control of MLCs. For example, the beam will be interlocked if a misplaced MLC leaf is detected, or if the MLCs are unintentionally used during electron mode.

• Many interlocks will be associated with the ion chambers that monitor the beam energy, flatness and symmetry. If any of these measured quantities drift out of the tolerance levels, then the interlock will be activated and irradiation will cease. Additionally, the beam is interlocked if the readings from the two ion chambers differ significantly.

• There are also interlocks for the correct selection of target, filter, scattering foil to match that at the control panel.

Abbreviations: MLC = Multileaf collimator.

Interlocking mechanism

(Radiotherapy) See Interlock; Interlocking device

Internal beam irradiation

(Nuclear Medicine) Cyclotrons are used to produce radionuclides by accelerating charged particles and then allowing them to hit a target. One way to irradiate the target is to insert it into the particle beam. This technique is called internal beam irradiation. Using this technique allows most of the accelerated particles to hit the target. The second alternative is to extract the particle beam by using either a stripping foil or an electrostatic deflector. This is called external beam irradiation. In general, external beam irradiation is the preferable method.

Related Articles: Cyclotron, External beam irradiation, Stripping foil


Internal conversion

(Radiation Protection) If an atomic nucleus is in an excited state \( \frac{1}{2} N \) * it releases the excess energy \( E_{exc} \), either by photon emission (gamma radiation) or internal conversion. During internal conversion the excess energy \( E_{exc} \), which would have been released by emission of gamma radiation, is instead absorbed by a bound electron, usually from a \( K \) or \( L \) atomic shell, and then that electron is ejected from the atom with a kinetic energy \( E_k \) equal to

\[
E_k = E_{exc} - E_b
\]

where \( E_b \) is the binding energy of the electron.

The vacancy created on the atomic shell after an internal conversion is filled by an electron from the higher shell (Figure I.28) with the subsequent emission of a characteristic x-ray.

The output of an internal conversion process can be estimated using a coefficient \( \alpha \), defined as the proportion of nuclear transitions which result in internal conversion, and can be compared to the proportion resulting in gamma emissions. This coefficient can

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**Image 1.26** In interleaved slice selection the slices are collected in a way that increase the distance between two slices collected after each other.

**Image 1.27** Interleaved k-space coverage for three slices using two shots in each slice.
be further refined to describe the rate of $K$-shell internal conversion – i.e. $\alpha_K$; similarly the rate for $L$-shell internal conversion – $\alpha_L$, and for $M$-shell internal conversion – $\alpha_M$.

**EXAMPLE:**

In the isomeric decay of $^{99m}$Tc to $^{99}$Tc, about 9% of the nuclear transitions result in $K$-shell internal conversion, 1.1% in $L$-shell, and 0.3% in $M$-shell. Then the total internal conversion coefficient is equal to 10.4%, whilst the remaining 89.6% of nuclear transitions result in gamma radiation emission.

**Related Articles:** Gamma rays, Isomeric transition (IT)


**Internal margin**

(Radiotherapy) A margin needs to be added to the clinical target volume (CTV) to form the planning target volume (PTV) to account for any positional errors from the planning information. The ICRU Report 62 divided this margin into the set-up margin (SM) and internal margin (IM), in order to separate out the contributory sources of positional error into physiological error and set up error, respectively. The IM compensates for physiological variation of the size and shape of the volume. It defines the difference between the CTV and the internal target volume (ITV), as shown in Figure I.29.

The IM incorporates both intra-fraction errors such as that due to respiration, and inter-fraction errors such as that due to weight gain/loss or digestive system changes. The latter inter-fraction error can be compensated to an extent by the use of portal imaging during each fraction.

The set-up margin (SM) is also required to account for errors in patient positioning and alignment of treatment beams, and the ITV plus this additional margin defines the extent of the PTV. The total required margin (SM + IM) can be quantified from the standard deviations ($\sigma$) of the associated probability distributions.

Equation I.2 shows combining margins for set up error and physiological (internal) error.

However this approach is not commonly used in practice to calculate margins. A more practical method is to decompose the margin into random and systematic components as explained in *Set-up error*.

**Abbreviations:** CTV = Clinical target volume, GTV = Gross tumour volume, ITV = Internal target volume, OAR = Organ at risk, PTV = Planning target volume and SM = Set up margin.

**Related Articles:** IM internal margin, Set up error, Portal imaging


**Internal photoelectric effect**

(Radiation Protection) An interaction between photonic radiation (ionising or non-ionising) and a material in which the absorption of a photon in a material results in the excitation of an electron from the valence band to the conduction band. This is of particular significance in the design of semiconductor radiation detectors, for example used in modern digital radiology.

**Related Articles:** Photoconductor effect, External photoelectric effect

**Internal radiation dosimetry**

(Radiation Protection) Internal radiation dosimetry employs biokinetic models and Monte Carlo mathematical techniques to determine the effective dose to a standard sized adult, and to children at
various ages, from the intended or accidental administration, ingestion or inhalation of a wide range of radionuclides in their common chemical forms. This includes radiopharmaceuticals used in nuclear medicine either for therapeutic or diagnostic purposes. An example of data available for use in nuclear medicine is the medical internal radiation dose (MIRD) tables and software. The biokinetic models the path of the radioactivity through the human body, describing how long the radioactivity remains in each organ or tissue. Knowledge of the total activity administered, and the biokinetic model, leads to the calculation of the absorbed dose to each organ over the time the activity remains in that organ (the committed absorbed dose), and thus to the effective dose over the time that the activity remains in the body (the committed effective dose).

**Related Articles:** Effective dose, Committed dose

**Internal reference point**

*(Radiotherapy, Brachytherapy)*

**Reference Points in Brachytherapy:** In classical brachytherapy, reference points were defined to specify dose to target and to calculate dose to organs at risk. Reference points were often defined in relation to the applicator(s), e.g. point A used to specify dose to the target in the Manchester system for treatment of cervix carcinoma. Radiographs were used to define reference points, and in the ICRU Report 38 standard reference points for rectum and bladder were defined and recommended for reporting in gynaecological brachytherapy.

The ‘simple’ ICRU bladder and rectum reference points, based on orthogonal radiographs, have been used extensively to characterise BT for cervix cancer in terms of maximum doses to these organs, in spite of their well known short comings. These ICRU reference points, as well as point A, are still used for reporting purposes to relate image-guided techniques to traditional planning methods.

The ICRU rectum and bladder point doses whose position can be determined from orthogonal radiographs often do not reflect the true maximum doses, as has been shown from studies using 3D image-based treatment planning. This is especially true for bladder where the dose determined using the ICRU reference point dose can significantly underestimate the maximum bladder dose value. A discussion of doses to organs at risk in relation to different side effects, maximum doses, doses to 1 cm³, and 2 cm³ volumes, etc. can be found in the following references.

As an example, **Figure 1.30** shows two orthogonal radiographs used to determine bladder and rectum ICRU points (slightly modified). A picture of the ring applicator itself and a standard dose distribution are included in the article *Intracavitary brachytherapy.*

1. Ring applicator including plastic spacers (not visible on the images)
   a. Diameter ring including spacer = 3.4 cm
   b. Intrauterine probe length = 4 cm
2. 7 mL contrast in Foley catheter balloon in bladder (Bl)
3. Special indicator with two rows of lead-shot in rectum (a, b)
4. Silver marker in cervix (elongated – for verification of appl. pos. between fractions)
5. Marker wires indicating source stop (dwell) positions inserted in both probe and ring
6. Calculations of total radiobiological effect (external beam radiation and brachytherapy) for organs at risk (dose restrictions) and target are made

**Abbreviations:** ICRU = International Commission on Radiation Units and Measurements.

**Related Articles:** Intracavitary brachytherapy, Volumetric prescribing – brachytherapy


**Internally deposited radionuclide**

*(Nuclear Medicine)* An internally deposited radionuclide is a radionuclide which is deposited in a dispersed form within the body. Examples include solutions, gases, dusts, and suspensions. These substances can enter the body by inhalation, ingestion, injection or percutaneous absorption.

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**FIGURE 1.30** HDR remote afterloading technique with ring applicator, using orthogonal x-ray films.
International Atomic Energy Agency (IAEA)

(General) The International Atomic Energy Agency (IAEA) was created in 1957 (‘Atoms for Peace’). It is an international organisation related to the United Nation (UN) as the worldwide centre for co-operation for the peaceful use of atomic energy. Medical applications are also included. In fact in its mandate as it is stated in article II of the IAEA statute is mentioned that: the agency shall accelerate and enlarge the contribution of atomic energy to health (recently this has been enlarged including also x-rays).

Typically, the IAEA is working in partnership with its Member States. The relation with the UN is regulated by a special agreement. The Agency reports annually to the UN General Assembly and, when appropriate, to the Security Council.

The Headquarter of the IAEA is based in Vienna, Austria, at the Vienna International Centre. Operational liaisons are located in other countries, namely Switzerland, the United States, Canada and Japan. There are also research centres and scientific laboratories run or supported by the agency, located in Austria, Monaco and Italy. In 2007 the regular budget of the IAEA was of 283, 611, 000 Euro (plus 80 million USD of voluntary contributions to the Technical Co-operation Fund), with a staff of 2200 multidisciplinary professional and support staff coming from 90 countries. Policy making bodies plan the work programme and the budget.

Within the aims of the agency, the programmes of work are tailored in order to support the various member states in their needs. In general there are three main areas of work: safety and security; science and technology; safeguard and verifications. Medical applications fall under the first two topics.

The agency is structured in departments. The activities are grouped as (1) technical cooperation (mainly cooperative projects in developing countries, providing training, specialised equipment and general support); (2) research and development (support to projects on critical problems related to radiation technologies in various fields including food, health, water and environment) and (3) energy and electricity (assessing of plans for energy needs including nuclear). Medical applications fall under the first two topics.

In particular, the division of human health, which belongs to nuclear science, is completely devoted to medical applications. The division is divided into four sections: (1) nuclear medicine; (2) proposed radiation biology and radiotherapy; (3) dosimetry and medical radiation physics and (4) nutritional and health-related environmental studies.

Hyperlinks: IAEA; http://www.iaea.org

International Commission on Non-Ionising Radiation Protection (ICNIRP)

(General) The principal focus and aim of ICNIRP is to disseminate information and give advice on health hazard regarding the exposure to non-ionising radiation; including the optical radiation (ultraviolet, visible and infrared, etc.), time-varying electric and magnetic fields, radiofrequency radiation and ultrasound.

Evaluations of the risk associated with the non-ionising radiation are usually carried out in collaboration with the World Health Organisation and the results are published as safety guidelines and when feasible include exposure limits. When the workers’ safety is involved, the collaboration is extended to the International Labour Organisation (ILO). The activity of ICNIRP is based on wide scientific expertise including medicine, epidemiology, biology, physics, dosimetry, etc. Therefore, in addition to the WHO and ILO, it is essential the partnership with the other related international organisations such as the International Radiation Protection Association (IRPA), the professional representative body for radiation professionals world-wide, the Institute for Electrical and Electronic Engineers (IEEE), the International Electrotechnical Commission (IEC), the European Commission (EC) and many others.

ICNIRP is legally registered in Germany and is a non-profit organisation. The modest income comes from IRPA, the German Environment Ministry, other governments and other sources with exemption of industries. Some income is provided by the sale of the publications. Many reports can be downloaded for free.

The structure of ICNIRP consists of a main commission and several standing committees. Each standing committee has its own work carried out in agreement with the main commission. Independent experts are called in to participate in the work of the different committees; they do not represent (either) their own country or their institute. They cannot be employed by industries. The expert work is given voluntarily.

Hyperlinks: www.icnirp.de; www.earthprint.com

International Commission on Radiological Protection (ICRP)

(Radiation Protection) The International Commission on Radiological Protection (ICRP) is an independent non-governmental organisation founded as the ‘International x-ray and Radium Protection Committee’ in 1928 by the Second International Congress of Radiology. The aim of the commission is to be the principal body to provide an appropriate international standard of protection for man without unduly limiting the benefit of the practices using ionising radiation. Although originally established to consider medical exposure to ionising radiation, in practice the scope has over the years widened from medical applications to include all exposures to ionising radiation from natural and artificial sources.

The commission is supported by a number of international organisations and governments. The main activity is to issue recommendations as an advisory body. The recommendations are made available to regulatory agencies at international, regional and national levels and they provide guidance on the fundamental principles on which radiation protections laws and regulations can be based. On the basis of these recommendations, other international organisations and regional/national authorities issue more detailed regulations and codes of practice.

Due to its historical origin, ICRP is an independent charity (not-for-profit organisation) registered in the United Kingdom, with a secretariat based in Stockholm, Sweden. It is composed of a main commission and five standing committees. The main commission consists of 12 members and a chairman, and the five standing committees cover: (1) radiation effects; (2) doses from radiation exposure; (3) radiation protection in medicine; (4) application of commission’s recommendations; and (5) protection of the environment. Each of these standing committee is chaired by a commission member, typically comprises 15–20 members, and they are collectively served by a small scientific secretariat.

ICRP may appoint working parties and task groups to aid in the preparation of reports. Working parties are formed by the standing committees (with the approval of the main commission) to develop ideas, and these will sometimes lead to the formation of task groups. Task groups usually contain specialists from outside the commission and are assigned responsibility for the preparation of draft reports. It remains the responsibility of the main commission to approve final reports for publication.

The commission issued its first report in 1928, which was subsequently numbered publication no.1 (the recommendations of publication no. 1 were adopted in 1958). Reports have been published on general and specialised topics related to radiation protection and are published as the Annals of the ICRP.
Financing of the ICRP comes mainly from voluntary contributions from international and national bodies; together with royalties from the commission's publications. Other forms of support come from member institutions.

**Related Article:** International Commission on Radiation Units and Measurements (ICRU)

**Hyperlink:** ICRU; http://www.icru.org

### International Commission on Radiation Units and Measurements (ICRU)

(Radiation Protection) The International Commission on Radiation Units and Measurements (ICRU) was founded, in 1925, by the 1st International Congress of Radiology. Since the beginning the principal objective has been the development of recommendations regarding: (1) quantities and units of ionising radiation and radioactivity; (2) procedures for the measurements and application of these quantities in radiation therapy, diagnostic radiology, radiation biology and industrial operations and (3) physical data needed in the application of the procedures. Therefore the ICRU collects and evaluates the latest data and information on the field of radiation protection and dosimetry and prepare recommendations for the most acceptable values and techniques for current use. It is essential to recognise accepted values for certain parameters in research work and application using ionising radiation The first task of the ICRU was that to devise a unit of ionising radiation that would make it possible to develop cancer treatment. Afterward many extensive sets of units and quantities have been developed, as applications became more complex and other fields of application were introduced.

The commission consists of 13 members selected for their scientific merits, regardless of nationalities, and is assisted by some twenty report committees which work on more ad hoc subjects. The committees (four to eight members) are appointed to produce draft documents and may also use external consultants. Two members of the commission ensure the connection between the committees and the commission.

ICRU is a not-for-profit organisation with a small secretariat. It is supported by a number of international organisations, intergovernmental bodies, foundations, industries, professional societies, etc. Circa one third of the budget comes from royalties for the publication of the recommendation.

Diagnostic imaging has become increasingly complex with high level of image elaboration, which requires common concepts, terminology and methodology. The ICRU has enlarged its programme, ranging from fundamental to practical aspects for all types of imaging techniques (including image quality) and has published reports on subjects related to modern diagnostic procedures related to x-ray and nuclear medicine investigations including also magnetic resonance imaging.

In order to make real progress in radiotherapy it is essential to compare results and methodology. Therefore a common language for reporting doses, techniques, fractionation schedule is crucial. The ICRU has put considerable effort in the definition of appropriate guidance levels (for defining tumours, target and planning volumes) and has given recommendations for the reporting of various treatments modalities.

Careful measurements are required for the protection of workers. The presence of ionising radiation and radioactive material in the environment influence the protection of the public and the environment. Due to diversity in routine and accidental exposures, international measurement standards are required for assessment of individual exposures and associated risks. All these aspects have been dealt with extensively.

The fundamental aspects of radiation science such as interaction of ionising radiation with matter are studied continuously. The related data are necessary in research on mechanisms of physical, chemical and biological changes following irradiation, and therefore also applications in medicine.

**Related Article:** International Commission on Radiological Protection (ICRP)

**Hyperlink:** ICRU; http://www.icru.org

### International Electrotechnical Commission (IEC)

(Radiation Protection) The International Electrotechnical Commission (IEC) was founded in 1906 with British scientist Lord Kelvin as its first president. IEC is the leading global organisation that prepares and publishes international standards for all electrical, electronic and related technologies. These serve as a basis for national standardisation and as references when drafting international tenders and contracts.

Through its members, the IEC promotes international cooperation on all questions of electrotechnical standardisation and related matters, such as the assessment of conformity to standards, in the fields of electricity, electronics and related technologies.

The IEC charter embraces all electrotechnologies including electronics, magnetics and electromagnetics, electroacoustics, multimedia, telecommunication and energy production and distribution, as well as associated general disciplines such as terminology and symbols, electromagnetic compatibility, measurement and performance, dependability, design and development, safety and the environment.

The commission's objectives are to

- Meet the requirements of the global market efficiently
- Ensure primacy and maximum world-wide use of its standards and conformity assessment schemes
- Assess and improve the quality of products and services covered by its standards
- Establish the conditions for the interoperability of complex systems
- Increase the efficiency of industrial processes
- Contribute to the improvement of human health and safety
- Contribute to the protection of the environment

### International Federation for Medical and Biological Engineering (IFMBE)

(General) IFMBE, the International Federation for Medical and Biological Engineering, is primarily a federation of national and transnational organisations. These organisations represent national interests in medical and biological engineering. The objectives of the IFMBE are scientific, technological, literary and educational. Within the field of medical, biological and clinical engineering IFMBE's aims is to encourage research and the application of knowledge, and to disseminate information and promote collaboration. IFMBE represents the interests of biomedical engineering profession in the UN and WHO as a non-governmental organisation.

Membership of all 60 affiliated societies from 55 countries constitutes Federation membership which is estimated to 120,000 individuals. IFMBE is affiliated with the International Union for Physical and Engineering Sciences in Medicine.

**Hyperlinks:** www.ifmbe.org

### International Organization for Medical Physics (IOMP)

(General) The International Organization for Medical Physics (IOMP) represents over 18,000 medical physicists worldwide and
has 82 affiliated national member organisations. The mission of IOMP is to advance medical physics practice worldwide by disseminating scientific and technical information, fostering the educational and professional development of medical physicists and promoting the highest quality medical services for patients.

Medical physics is a branch of applied physics, pursued by medical physicists, which uses scientific principles, methods and techniques in practice and research for the prevention, diagnosis and treatment of human diseases with a specific goal of improving human health and well-being.

A medical physicist is a person, qualified with a university degree majoring in physics with specialised education and training in the concepts and techniques of applying physics in medicine and healthcare.

The International Organization for Medical Physics (IOMP) was formed in 1963 and has the following objectives:

1. To organise international cooperation in medical physics and to promote communication between the various branches of medical physics and allied subjects.
2. To contribute to the advancement of medical physics in all its aspects.
3. To advise on the formation of national organisations of medical physics in those countries which lack such organisations, and also the possible formation of national committees in those countries where there is more than one medical physics organisation.

In 1980 IOMP, together with the International Federation of Medical and Biological Engineering (IFMBE), formed the International Union of Physical and Engineering Sciences in Medicine (IUPESM) as an umbrella body. The IUPESM became a member of the International Council of Scientific Unions (ICSU) in 1999.

In 2005 IOMP formed a relationship with the International Union of Pure and Applied Physics (IUPAP), to strengthen collaboration of medical physicists with other physicists with similar interests. Within the IUPAP organisation, IOMP is recognised as an Affiliated Commission and is referred to as the International Commission on Medical Physics (ICOMMP). Within IOMP, the International Commission on Medical Physics (ICOMMP) Committee facilitates collaboration of the medical physicists with the physicists who have the similar academic interests in research and education.

IOMP has three categories of membership:

1. Individual members: These are the members of national organisations associated with adhering national bodies.
2. Adhering bodies: National medical physics organisations or societies.
3. Affiliated organisations: Regional organisations and corporate members.

IOMP undertakes a broad range of activities – scientific, education and training and dealing with professional matters. There is particular emphasis in supporting the development of medical physics in developing countries.

The IOMP has two major series of meetings – the triennial World Congresses of Medical Physics and Biomedical Engineering, organised though IUPESM, and the International Conferences of Medical Physics held, between World Congresses and more regional in nature. Individual sessions or seminars are also organised at other international meetings.

**Hyperlinks:** IOMP; www.iomp.org

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**International Radiation Protection Association (IRPA)**

(Radiation) The real beginning of the International Radiation Protection Association (IRPA) starts as an initiative of the American Health Physics Society to form a committee to deal with radiation protection problems at international level. After several years of work and a much growing interest in this field, it was decided to create an international health physics society to associate national societies. The first pro tempore general assembly was held in Paris in 1964, with the participation of 45 delegates from 15 national health physics or radiation protection societies. At the Paris meeting a constitution was adopted and the primary objectives were decided.

The primary objective of IRPA is to provide a medium whereby radiation protection professionals from all countries may communicate with each other and in this way support the development and advance of radiation protection. Other knowledge relevant fields and scientific aspects related to radiation protection such as medicine, engineering, technology, law, etc. are also included. This is in order to provide for a better protection from the hazardous effects of ionising radiation and at the same time facilitate and optimise its beneficial uses.

Other objectives of IRPA include (1) encourage the formation of radiation protection societies with the purpose to improve international co-operation, (2) support international meetings, (3) encourage international publications, (4) encourage research and education in radiation protection and related topics and (5) encourage the establishment and continuous review of universally accepted radiation protection standards and recommendations.

The main organisational structure of IRPA consists of the associated societies, the general assembly and the executive council, with a secretariat and a treasury. In addition there are five commissions and committee dealing with various topics. The executive functions of IRPA are performed by the officers upon approval of the executive council. An international conference is held every 4 years. The conference represents a major forum for discussion in the field of radiation protection.

**Hyperlink:** IRPA; www.IRPA.org

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**International Union for Physical and Engineering Sciences in Medicine (IUPESM)**

(General) International Union for Physical and Engineering Sciences in Medicine (IUPESM) – the principal objective of IUPESM is to contribute to the advancement of physical and engineering sciences in medicine for the benefit and well being of humanity. Its objectives include organising international cooperation and promoting communication among those engaged in health-care science and technology, coordinating activities of mutual interest to engineering and physical science within the health care field, including international and regional scientific conferences, seminars, working groups, regional support programs and scientific and technical publications and representing the professional interests and views of engineers and physical scientists in the health-care community.

The IUPESM represents the combined efforts of more than 40,000 medical physicists and biomedical engineers working on the physical and engineering science of medicine. The IUPESM is the umbrella organisation of the International Federation for Medical and Biological Engineering (IFMBE) and International Organization for Medical Physics (IOMP).

IUPESM is a full member of International Council of Science Union (ICSU).

IUPESM is sponsoring and coordinating the triennial ‘World Congress for Medical Physics and Biomedical Engineering’. (www.iupesm.org)
Interpolation

(Atomic, Nuclear Medicine) Interpolation refers to the method of calculating a value in a point in-between two known points in a discrete data set. The most common form of interpolation is linear interpolation where all the values are estimated from a straight line between the two known points. Interpolation can also be seen as a specific case of curve fitting where the curve must pass through each known data point.

 Interruption of treatment

(Radiotherapy) Generally, radiotherapy treatment is delivered once a day, 5 days a week for up to 8 weeks. Many of the regimens currently in use have developed as a result of expediency rather than from radiobiological principles, evolving to accommodate the standard 5 day working week. They are considered to compensate empirically for tumour repopulation during the non-treatment weekend breaks but the addition of further interruptions to the planned treatment schedule may result in prolongation of the overall treatment time. There is extensive evidence that the prolongation of treatment increases the risk of local recurrence in a wide range of fast-growing tumours and it should therefore be avoided. However, there are occasions where unforeseen interruptions occur such as machine breakdown or illness of the patient. There have been few studies on the effect of treatment prolongation on patients with slow-growing tumours but it is expected that any interruption to a radiotherapy schedule may affect outcome.

In the United Kingdom, the Royal College of Radiologists have addressed the issue of treatment interruptions and published guidance for clinical practice. They recommend that all radiotherapy departments establish robust systems of service planning to cope with predictable and unpredictable interruptions to normal treatment including working across bank holidays and the provision of adequate resources in terms of machines, staff and training. Where an interruption does occur, they recommend one of the following courses of action is taken, listed in order of preference, with priority given to patients with rapidly growing tumours being treated with radical intent:

1. If due to machine breakdown, transfer of patient to a matched linear accelerator on the day of interruption.
2. Missed weekday treatment fraction to be delivered at the weekend.
3. Patient treated twice daily with a minimum of 6 h between fractions, preferably towards the end of the week to allow repair of sub-lethal normal tissue damage over the ensuing weekend. Twice daily treatment is not recommended when fraction size is significantly greater than 2.2 Gy.
4. Use of biologically equivalent dose (BED) calculations to derive an alternative schedule with a modified number of treatment fractions in order to complete the radiotherapy course in the original planned overall time.
5. The addition of extra treatment fractions where it is not possible to maintain the original planned overall time even with compensation.

The compensation options available will depend on when during a course of treatment an interruption occurs. The later this is, the less likely it is that the original overall treatment time can be maintained. It should be noted that in cases 3–5, the BED to normal tissues may be higher than that for the original treatment prescription. For cases 4 and 5, a compromise will usually be required between reducing the BED to the tumour and increasing the BED to normal tissues. Compensation requiring the use of radiobiological-based calculations should only be adopted when other methods of compensation can not be applied since assumptions need to be made for parameter values and should only be carried out by appropriately trained physicists or clinicians. A discussion of these methods can be found in the paper by Dale et al. along with a number of practical examples. These examples are reproduced in Appendix B of the Royal College of Radiologists’ report.

 Related Articles: Biological effective dose (BED), Fractionation, Repair, Repopulation, The 5 R’s of radiobiology, Tumour control probability.


Intersource shielding

(Radiotherapy, Brachytherapy)

Intersource Shielding – Intersource Effect: The source models used in treatment planning systems describe the dose distribution around the source in a homogeneous ‘infinite’ medium. The dose distribution calculations in the treatment planning systems include no corrections for heterogeneities of any kind (2009). Consequently, when many sources are implanted, as in seed implantations for prostate cancer, the seeds will shield each other. Thus, the total dose is overestimated when calculated as a superposition of the dose distributions from the individual sources. An estimation of the reduction of the dose to the prostate shows that a reduction of several per cent is possible.

It is to be noted, that all treatment planning systems today calculate dose this way. Thus, when dose calculation algorithms developments include heterogeneity corrections, in particular intersource effects, this will give a more accurate calculation of given doses. In addition it will be possible to translate existing dose levels used for specification of prostate seed implants to new and more accurate values (Figure I.31).

For an afterloading system with one stepping source, there are no intersource effects.

 Related Articles: Treatment planning systems – brachytherapy, Source models

Interstitial

(Radiotherapy, Brachytherapy)

FIGURE I.31 Seed implant, 125I-seeds, intersource effects.
Placement of Sources in Brachytherapy: In brachytherapy, applicators and sources are placed inside or close to the target volume using different techniques.

Interstitial Technique:

- Applicators/sources are placed inside the target volume (you make the 'cavities' yourself)
- Interstitium (Latin), meaning 'a space between', inter (Latin), meaning between

See Interstitial brachytherapy

Related Articles: Brachytherapy, Interstitial brachytherapy, Intracavitary brachytherapy

Interstitial brachytherapy (Radiotherapy, Brachytherapy) Placement of Sources in Brachytherapy: In brachytherapy, applicators and sources are placed inside or close to the target volume using the following techniques:

1. Intracavitary technique
   a. Applicators/sources are placed in existing cavities, inside or close to the target volume
   b. The term is derived from intra (Latin), meaning ‘within, inside’
   c. This technique is traditionally used for gynaecologic tumours
      i. Applicators/sources of different design are placed in the vagina and/or the uterine cavity
      ii. There are a large number of applicator types designed for brachytherapy of cancer of the cervix
      iii. For tumours in the vaginal mucosa, vaginal cylinders are used
   d. A subcategory of this technique is the intraluminal technique, lumen (Latin), meaning 'passage within a tubular organ', for example intraluminal brachytherapy of the bronchus – endobronchial brachytherapy, where applicator/sources are placed in the bronchus (endon [Greek], meaning ‘within’)
   e. As placement of sources is restricted to existing cavities, dose gradients are larger in intracavitary brachytherapy than in interstitial techniques (see point 2), where sources are distributed ‘more freely’
   f. The intracavitary technique is applicable only for temporary implants

2. Interstitial technique
   a. Applicators/sources are placed inside the target volume (you make the 'cavities' yourself)
   b. Interstitium (Latin), meaning 'a space between', inter (Latin), meaning between
   c. Applicators used are
      i. Needles, guided by a template; for prostate cancer (see following section), mammary cancer, etc.
      ii. Catheters; e.g. for head and neck tumours
   d. This technique is traditionally used for other sites than the gynaecological region
   e. It is applicable for both permanent and temporary implants

3. Surface applications
   a. In principle a border case of intracavitary technique

Brachytherapy Applications – Interstitial Techniques:

EXAMPLE 1

Prostate cancer: Interstitial interactive permanent implant of $^{125}$I seeds, very low dose rate brachytherapy

The procedure is ultrasound guided, using a template for needle positioning, and a dedicated treatment planning system to plan and follow the implantation (see also Permanent implant).

Figure I.32 shows the dose distribution as planned with ideal needle positions.

During the implantation, needle (and seed) positions in the treatment planning system are adjusted to the real needle (and seed) positions interactively. The resulting dose distribution, which is continuously updated, is followed closely during the implant procedure, and extra seeds can be added as desired.

FIGURE I.32 (See colour insert.) Dose distribution as planned with ideal needle positions, using a dedicated treatment planning system (VariSeed, Varian).
EXAMPLE 2

Prostate cancer: Interstitial interactive temporary implant using high dose rate brachytherapy

The procedure is ultrasound guided, using a template for needle positioning, and a dedicated treatment planning system to plan and follow the implantation (see also Temporary implant).

These needles are closed, to ensure integrity of the high intensity 192Ir source of the remote controlled after loading unit used for the treatment.

A preliminary dose distribution with ideal needle positions is the starting point. After insertion of all the needles, the needle positions are adjusted in the treatment planning system, a final plan is made (see Figure I.33) and approved, the patient is treated, and the needles are taken out.

Related Articles: Brachytherapy, Intracavitary brachytherapy, Permanent implant, Temporary implant, Treatment planning systems – brachytherapy

Interstitial implant

(Radiotherapy, Brachytherapy) Placement of Sources in Brachytherapy: See Interstitial brachytherapy

Related Articles: Brachytherapy, Interstitial brachytherapy, Intracavitary brachytherapy, Permanent implant, Temporary implant

Interstitial therapy

(Radiotherapy, Brachytherapy) Placement of Sources in Brachytherapy: See Interstitial brachytherapy

Related Articles: Brachytherapy, Interstitial brachytherapy, Intracavitary brachytherapy

Interventional MRI

(Magnetic Resonance) Interventional MRI (iMRI) refers to the performance of invasive clinical procedures under MRI guidance. This is currently a small field with a diverse range of applications, including minor interventional radiology procedures (such as breast biopsies), minimally invasive procedures (such as tumour ablation and interventional cardiology) (Figure I.34) and even open neurosurgery.

Interventional MRI is developing as an alternative to conventional x-ray-guided interventional procedures. Advantages include lack of ionising radiation exposure to patients and staff, better soft tissue visualisation and contrast, multi-planar imaging capability, and the ability to collect images depicting physiological or mechanical information.

There are also significant disadvantages. Conventional MRI systems allow little access to the patient, and although open systems are available these are usually low field systems with significantly compromised image quality, and most still do not offer ideal access. Furthermore, many medical devices designed for conventional interventional use are poorly visualised on MRI, or else result in image degradation on account of their magnetic properties and/or electrical conductivity.

Many devices, i.e. tools or implants, also present safety concerns in the MRI environment, so care has to be taken to use only MR-safe devices. Special versions of some interventional devices have been developed for MRI use, but as yet there is little commercial motivation for this. There are also wider safety concerns relating to the involvement of members of staff, such as surgeons and nurses, who may not be familiar with MRI safety issues, so rigorous safety procedures and training are prerequisite.

Interventional MRI has required, and has fuelled, the development of real-time imaging techniques and rapid reconstruction algorithms, in order to provide the multi-frame per second temporal resolution needed for interventional guidance.

Related Article: Real-time imaging


FIGURE I.33 (See colour insert.) The final dose distribution after needle position adjustment (BrachyVision, Varian).

FIGURE I.34 Paediatric cardiologist performing MR-guided cardiac catheterisation using a 1.5 T scanner.

**Intracavitary**

(Radiotherapy, Brachytherapy) *Placement of Sources in Brachytherapy*: In brachytherapy, applicators and sources are placed inside or close to the target volume using different techniques.  

**Intracavitary Technique**: Applicators/sources are placed in existing cavities, inside or close to the target volume. Intra (Latin), meaning within, inside

See also Intracavitary brachytherapy

**Related Articles**: Brachytherapy, Intracavitary brachytherapy, Interstitial brachytherapy

**Intracavitary**

(Ultrasound) Intracavitary (also referred to as intracavity) transducers are designed to image from within a body cavity. These allow placement of the transducer elements near to the target tissue and enable improved image quality.  

Commonly used transducers are

1. Endorectal transducers, used to image the prostate gland  
2. Endovaginal transducers, used in gynaecology examinations and early pregnancy  
3. Transoesophageal transducers for cardiac imaging

Various designs are used, usually with tightly curved or phased arrays. Arrays can be aligned along the axis of the transducer and/or transversely across. The use of intracavitary transducers allows high frequencies to be used. The available aperture is limited by the physical restrictions necessary to allow penetration into the body cavities (Figure I.35).

**Intracavitary brachytherapy**

(Radiotherapy, Brachytherapy) *Placement of Sources in Brachytherapy*: In brachytherapy, applicators and sources are placed inside or close to the target volume using the following techniques (Figures I.36 through I.40):

1. Intracavitary technique
   a. Applicators/sources are placed in existing cavities, inside or close to the target volume  
   b. The term is derived from intra (Latin), meaning ‘within, inside’  
   c. Traditionally used for gynaecologic tumours

**FIGURE I.35** Intracavitary transducer designed for endovaginal imaging. The tightly curved array (left) images forwards. The handle (R) allows the probe to be manoeuvred.

**FIGURE I.36** Cervical cancer – Lund old Radium applicator, coupled box and probe.

**FIGURE I.37** Cervical cancer – Lund newer ring applicator, for high dose rate treatments, remote controlled afterloading (Varian).

**FIGURE I.38** Vaginal cylinder, Lund, high dose rate (Varian).
Intracavitary brachytherapy

Applicators/sources of different design are placed in the vagina and/or the uterine cavity.

There are a large number of applicator types designed for brachytherapy of cancer of the cervix.

For tumours in the vaginal mucosa, vaginal cylinders are used.

Subcategory: intraluminal technique, lumen (Latin), meaning 'passage within a tubular organ'.

Example: intraluminal brachytherapy of the bronchus – endobronchial brachytherapy, where applicator/sources are placed in the bronchus (endon [Greek], meaning 'within').

As placement of sources is restricted to existing cavities, dose gradients are larger in intracavitary brachytherapy than in interstitial techniques, where sources are distributed 'more freely'.

Applicable for both permanent and temporary implants.

Interstitial technique

Applicators/sources are placed inside the target volume (you make the 'cavities' yourself).

Interstitium (Latin), meaning 'a space between', inter (Latin), meaning 'between'.

Applicators used are:

i. Needles, guided by a template; for prostate and mammary cancer, etc.

ii. Catheters, e.g. for head and neck tumours.

Traditionally used for other sites than the gynaecological region.

Applicable for both permanent and temporary implants.

Surface applications

In principle a border case of intracavitary technique.
Intracavitary therapy

Intraoperative radiotherapy

Related Articles: Brachytherapy, Intracavitary brachytherapy, Interstitial brachytherapy

Intraluminal brachytherapy

(Radiotherapy, Brachytherapy) Placement of Sources in Brachytherapy: In brachytherapy, applicators and sources are placed inside or close to the target volume using different techniques.

Intracavitary Technique:

- Applicators/sources are placed in existing cavities, inside or close to the target volume
- Intra (Latin), meaning ‘within, inside’
- Subcategory; intraluminal technique, lumen (Latin), meaning ‘passage within a tubular organ’. For example intraluminal brachytherapy of the bronchus, also denoted ‘endobronchial’ brachytherapy (endon [Greek], meaning ‘within’), where applicator/sources are placed in the bronchus

Isocentric films, high dose rate intraluminal brachytherapy of a bronchial tumour, target length 5cm, 9 stop positions, step size 0.5cm, dose specified 10mm from the stop positions (Figures I.41 through I.43).

See Intracavitary brachytherapy

Related Articles: Brachytherapy, Intracavitary brachytherapy, Interstitial brachytherapy

Intraoperative radiotherapy

(Radiotherapy) Intraoperative radiotherapy (IORT) involves the delivery of a radiotherapy dose at time of surgery for local control of cancer. The main example of this is during surgery for breast cancer. The radiotherapy is often delivered using a small size x-band linac with a range of electron energies, typically between 4 and 12MeV. Another approach is to deliver a lower energy x-ray beam (50keV) using a spherical applicator positioned at the end of an electron accelerator. An example is the TARGIT system.

Abbreviation: IORT = Intraoperative radiotherapy.

FIGURE I.40 (continued) (See colour insert.) Example of dose distribution for the two Lund cervix applicators. (b), (c) and (d) Dose distributions for three orthogonal sections through the ring applicator. Ring diameter with spacer 4.1 cm, intrauterine probe length 4 cm. Dose specification; 5 Gy, 2 mm outside the spacer ventrally and dorsally. Dose at point A 5 Gy, dose to organs at risk (rectum and bladder), nominally 5 mm outside the spacer; 77% of specification dose.

Related Articles: Brachytherapy, Interstitial brachytherapy, Temporary implant

Intracavitary therapy

(Radiotherapy, Brachytherapy) Placement of sources in Brachytherapy: See Intracavitary brachytherapy

FIGURE I.41 AP film.
Intravascular irradiation

(Radiotherapy, Brachytherapy) Placement of Sources in Brachytherapy: In brachytherapy, applicators and sources are placed inside or close to the target volume using the intracavitary and interstitial techniques.

Intracavitary Technique:

- Applicators/sources are placed in existing cavities, inside or close to the target volume
- Intra (Latin), meaning ‘within, inside’
- Subcategory; intraluminal technique, lumen (Latin), meaning ‘passage within a tubular organ’

Example: intraluminal brachytherapy of the bronchus, also denoted ‘endobronchial’ brachytherapy (endon [Greek], meaning ‘within’), where applicator/sources are placed in the bronchus.

An intraluminal technique is also used in intravascular irradiation, also denoted intravascular brachytherapy or endovascular brachytherapy. Sources are placed inside the lumen of an artery, and the intention of the treatment is to prevent vascular restenosis (i.e. to prevent the vessel from ‘becoming narrow again’).

Both gamma-emitting sources, such as the traditional iridium-192 source, and beta-emitting sources, such as strontium-90 and phosphorus-32, have been used.

One specific physics challenge in intravascular brachytherapy is the dosimetry very close to the sources (distance <2 mm); a region of comparatively smaller interest in ‘traditional’ brachytherapy dosimetry, where generally distances of the order of 5–10 mm are considered.

Some of the earlier randomised trials in intravascular brachytherapy were performed in the 1990s; covering restenotic lesions in the superficial femoral artery and stenting versus stenting plus brachytherapy of coronary arteries. The results were encouraging and the use especially of intracoronary brachytherapy to control coronary restenosis grew rapidly. But, drug coated stents became available in the early 2000s. These coated stents were easy to use and gave very good treatment results. As a consequence, the use of brachytherapy in the treatment of cardiac restenosis disappeared quickly in favour of the treatment where no radiation was used. (For a detailed description of intravascular brachytherapy, stenting, vascular restenosis, etc., the reader is referred to existing literature.)

Related Articles: Brachytherapy, Intracavitary brachytherapy, Interstitial brachytherapy

Intravascular ultrasound (IVUS)

(Ultrasound) There are several reasons to visualise arteries and veins from within. The close proximity to the walls allows the use of high frequency with consequent high resolution, allowing detailed examination of the vessel wall. This includes information concerning the extent and acoustic properties of plaques and the properties of the vessel walls and histological layers since the scanning range is up to 2 cm. This information is not available from angiography. In addition, IVUS allows the assessment of stenosis with the measurement of luminal area before and after atherectomy (Hedrick et al., 1995).

Frequencies of 20–50 MHz are typically used for intravascular ultrasound, with a corresponding wavelength of approximately 0.03 mm.

The obvious concern is that IVUS is invasive, but Wui K. Chong (in Lees and Edward, 1996) states that apart from a few cases of transient coronary artery spasm, there have been no other complications reported from the use of IVUS and that the technology is simple and safe.


Intravoxel incoherent motion (IVIM)

(Magnetic Resonance) The terminology intravoxel incoherent motion (IVIM) refers to the effect related to phase dispersion present within voxels subject to random microscopic motion. In contrast, intravoxel coherent motion (IVCM) is contributed to constant flow within a voxel and where a phase angle can be determined and used to estimate the speed of, e.g. flowing blood. The concept of IVIM was introduced by Denis Le Bihan in the late 1980s in order to characterise random motion in vivo. IVIM includes effects from both the water self-diffusion and the microcirculation of blood in the capillary bed, i.e. perfusion. As detected by a diffusion encoding pulse sequence, these two types of motions are entirely equivalent, with the only difference that the perfusion motion is faster. The diffusion coefficient resulting from the perfusion fraction is sometimes denoted the pseudo-diffusion. The effects from the perfusion motion can be identified in the early part of the signal versus diffusion-sensitivity decay curve, but due to the speed of the pseudo-diffusion, this part is attenuated already at diffusion sensitivities on the order of 300 s/mm². In conventional diffusion
imaging where the apparent diffusion coefficient (ADC) is estimated based on measurements performed with diffusion sensitivity zero \( (b = 0) \) and \( b = 1000–1500\text{s/mm}^2 \), the ADC will be overestimated due to the perfusion related signal decay. In cerebral imaging this overestimation will be the greatest in the grey matter.

**Related Articles:** Apparent diffusion coefficient (ADC), b-value

**Intrinsic amplifier noise**

*(General)* All electronic amplifiers introduce some additional electronic noise to the signal being amplified. The common sources of noise are as follows:

**Thermal Noise (Johnson–Nyquist Noise):** This is caused by the thermally generated movement of electrons in conductors and defined mathematically as

\[ V_n = \sqrt{4kTRB} \]

where

- The noise voltage \( V_n \) is given as RMS mean voltage
- \( R \) is the resistance of the conductor
- \( T \) is the absolute temperature
- \( B \) is the bandwidth in Hz of the noise
- \( k \) is Boltzmann’s constant \( (1.38 \times 10^{-23}\text{m}^2\text{kg}/\text{s}^2/\text{K}) \)

Thermal noise may be minimised by cooling the amplifier, minimising resistances by keeping all impedances low, using reactive components where possible (\( C \) and \( L \) have no significant thermal noise), and limiting bandwidth by appropriate filtering.

**Shot noise:** This additional noise is generated in any circuit where current is flowing, due to the statistical fluctuation in current as it is made up from many individual charges moving about:

\[ I_n = \sqrt{2qIB} \]

where

- The noise current \( I_n \) is given as an RMS value
- \( I \) is the mean current flowing
- \( B \) is the bandwidth of the noise
- \( q \) is the unit electron charge \( (1.6 \times 10^{-19}) \)

As the current increases, the proportion of noise to signal decreases – thus where possible shot noise can be minimised by avoiding the use of low currents.

**Flicker (I/f) Noise:** This is a further source of noise associated with the flow of current through real components. It is significantly greater at low frequencies, with noise power being inversely proportional to frequency. This noise becomes dominant over other noise sources at frequencies below a few 100 Hz. Flicker noise can be minimised by selecting components of superior quality.

**Noise Figure (NF):** Amplifiers, especially RF amplifiers, are commonly given a figure of merit reflecting the additional noise that the amplifier introduces:

\[ NF_{\text{in}} = 10\log\left(\frac{\text{SNR}_{\text{in}}}{\text{SNR}_{\text{out}}}\right) \]

In practice this comparison of the signal-to-noise ratios is a gain-independent guide to how well the amplifier performs (NF = 0 dB being perfect).

In practice these sources of noise are small (nV to µV) and interference from other electronic sources will be much greater that intrinsic noise in most applications.

**Abbreviation:** NF = Noise figure.


**Intrinsic efficiency**

*(Radiation Protection)* The intrinsic efficiency \( \varepsilon_{\text{int}} \) of a radiation detector is defined as a quotient of the number of recorded pulses \( N_{\text{rec}} \) to the number of radiation quanta (particles or photons) \( N_{\text{rad}} \) incident on detector:

\[ \varepsilon_{\text{int}} = \frac{N_{\text{rec}}}{N_{\text{rad}}} \]

The intrinsic efficiency depends on the detector material, its thickness, and the type and energy of the incident radiation.

The absolute detection efficiency \( \varepsilon_{\text{abs}} \) is expressed as a ratio of the number of recorded pulses \( N_{\text{rec}} \) to the number of radiation quanta emitted \( N_{\text{em}} \) by a radiation point source:

\[ \varepsilon_{\text{abs}} = \frac{N_{\text{rec}}}{N_{\text{em}}} \]

For an isotropic source, the relationship between intrinsic and absolute detection efficiency is therefore:

\[ \varepsilon_{\text{int}} = \varepsilon_{\text{abs}} \left( \frac{4\pi}{\Omega} \right) \]

where \( \Omega \) is a solid angle (in steradians – sr) subtended by the detector from the position of the radiation point source.

**Related Article:** Detection efficiency


**Intrinsic flood field uniformity**

*(Nuclear Medicine)* Scintillation camera uniformity measured without a collimator using a point source (providing a uniform irradiation of the detector surface). Two different parameters are of interest; integral and differential uniformity. **Integral uniformity** is defined as a measure of the maximum pixel count deviation in the central field of view (CFOV) and useful field of view (UFOV). This is calculated by dividing the difference between the maximum and minimum pixel value by their sum and then multiplying by 100:

\[ \text{Integral uniformity} = \pm 100 \times \left( \frac{\text{max} – \text{min}}{\text{max} + \text{min}} \right) \quad (I.3) \]

The differential uniformity is defined as the maximum deviation between five contiguous pixels:

\[ \text{Differential uniformity} = \pm 100 \times \left( \frac{\text{max} – \text{min}}{\text{max} + \text{min}} \right) \quad (I.4) \]

Each row and column in the image is treated as separate slices. A set of five contiguous pixels are selected in the slice, the maximum and minimum value is identified and used to calculate the differential uniformity for that particular set. This procedure is continued for each row and column within the specified area (e.g. CFOV/ UFOV). The maximum differential uniformity is then identified.
Inverse Fourier transform

(General) The Fourier transform is a very common method to express signals or images as periodical sine and cosine functions with an amplitude and a frequency. The definition of the Fourier transform $F$ of a function $f$ is as follows:

$$ F(\omega) = \int_{-\infty}^{\infty} f(t)e^{-j\omega t} dt $$

The inverse Fourier transform will then be

$$ f(t) = \frac{1}{2\pi} \int_{-\infty}^{\infty} F(\omega)e^{j\omega t} d\omega $$

Inverse radiotherapy planning

(Radiotherapy) Radiotherapy treatment planning involves choosing the beam directions, field shapes and intensities for the patient’s beam delivery. This is usually done using a CT scan of the patient and a model of dose deposition in the patient by the beam. In complex treatments, such as IMRT, this is often achieved using an automated, optimisation process. It is called inverse planning in that the planner specifies the desired dose distribution and the optimisation system computes the optimum beam distribution to deliver a plan that matches the prescribed dose distribution as closely as possible; thus, it is the opposite of normal, interactive forward planning. The desired dose distribution is usually specified in terms of the desired dose to the target (including allowed variation from that dose) and dose limits to normal tissues, plus importance factors for the various tissue regions. In some cases constraints on the dose volume histogram or desired calculated TCP and NTCP values may be specified. The optimisation is often carried out using a standard least squares gradient descent method or simulated annealing. Upon satisfactory calculation of the required fluence maps of the IMRT beams, the treatment planning system can then generate for each of the beams a set of multi-leaf collimator (MLC) motion sequence files (dynamic delivery), or a series sub fields (step and shoot delivery), which can be transferred to the MLC controller of the treatment machine for delivery of the treatment beams.

Abbreviation: CT = Computed tomography.

Related Articles: Treatment planning, Interactive planning, Dose volume histogram

Inverse square law correction

(Radiotherapy) It is reasonable to assume that the source of x-ray photons is a point source, and therefore the beam that is produced is divergent in nature. Therefore assuming a square field of side $a$ (with an area $A = a^2$) is produced at a distance $d_a$ from the source and a square field of side $b$ (area $B = b^2$) is produced at a distance $d_b$, the two can be related as follows:

$$ \frac{a}{b} = \frac{d_a}{d_b} $$

Assuming the source produces a total of $N$ x-ray photons with a fluence of $\Phi_A$ at $d_a$, and fluence $\Phi_B$ at $d_b$ from the source. Then the two can be related as follows:

$$ N = \Phi_A A = \Phi_B B $$

and

$$ \frac{\Phi_A}{\Phi_B} = \frac{B}{A} = \frac{b^2}{a^2} = \frac{d_b^2}{d_a^2} $$

So the fluence is inversely proportional to the square of the distance from the source, and by extension the exposure, air kerma in air and dose will likewise be inversely proportional to the square of the distance from the source. It is necessary to allow for this inverse square law correction when performing calculations at different distances from those where the dose calibration is performed. For example, typically the linac output is quoted at the depth of maximum dose ($z_{max}$) for a square field of side 10 cm measured at the nominal SSD of the unit (typically 100 cm). Therefore for isocentric treatment calculations the dose must be corrected using an inverse square correction factor (unless the linac is actually calibrated at isocentre) as shown here:

$$ ISLCorr = \left( \frac{SSD + z_{max}}{SSD} \right)^2 $$

Inversion recovery

(Magnetic Resonance) In a pulse sequence starting with a 90° excitation pulse, such as the spin echo (SE, 90°–180°) and the fast spin echo (FSE, 90°–180°–180°–180°–…), $T_1$ contrast is obtained by adjusting the sequence repetition time TR so that object parts with different $T_1$ values, due to different longitudinal recovery during TR, obtain different signal values. The signal ranges between 0 (TR = 0) and $S_{max}$ ~ PD (TR → ∞) if transversal relaxation effects are neglected. This signal range, creating $T_1$ contrast, can be increased by introducing an inversion pulse (i.e. a 180° pulse), thereby inverting the direction of $M_x$ prior to the 90° pulse used to flip the magnetisation into the transversal plane (Figure I.44). This experiment, denoted inversion recovery, was early shown to give good clinical (morphological) images (1). In order to prevent susceptibility-induced dephasing, a full SE or FSE sequence can be executed after the inversion pulse instead of only a 90° pulse. The obvious drawback with spin-echo-based IR sequences is long acquisition times since commonly high TR values are chosen. A way of overcoming this problem is to create a fast inversion recovery sequence type, often also denoted turbo inversion recovery. In this sequence type, the series of RF pulses during one repetition are 180°–90°–180°–180°–180°–… Fast IR sequences have opened the possibility to perform the IR experiment.
within reasonable acquisition times, since in this case $T_{eq}$ will be reduced by the echo train length (ETL) factor as in TSE.

It can be shown that the IR signal is approximated by (neglecting transversal relaxation)

$$S_{IR} \approx PD \left(1 - 2e^{-T_{1}/T}\right) - e^{-TR/T_1}$$

where TI (inversion time) is the time between the inversion pulse and the 90° pulse. The signal range in this case, assuming $TR≫T_1$, $TR>TI$, goes between $-S_{max} ≈ PD(TI = 0)$ and $S_{max} ≈ PD(TI → ∞)$. Hence the dynamic signal range for IR will be about twice the range for SE, allowing for increased TI contrast (Figure 3). Another feature of interest with the IR pulse sequence is that the signal as a function of TI is nulled at a TI which depends upon TR and $T_1$, indicating that object parts with specific and known $T_1$ values can be nulled for a properly chosen TI. For very long TR values $T_{1,nul} = 0.69 T_1$. With respect to image contrast, the IR image can be visualised in two different ways: Using phase-sensitive reconstruction the greyscale can be adjusted for negative as well as positive signal values, so that negative signal values will be dark, zero signal gray and positive signal values bright. This type of display will yield a characteristic gray background outside the object (zero signal) and occasional ‘background-like’ parts inside the object for object parts with nulled signal. Another representation is provided by magnitude reconstruction, where absolute signal values are displayed. For object parts with nulled signal. Another representation is provided by magnitude reconstruction, where absolute signal values are displayed. For negative as well as positive signal values, so that negative signal values will be dark, zero signal gray and positive signal values bright. (c) Another representation is provided by magnitude reconstruction, where absolute signal values are displayed.

**Inversion time delay (TI delay)**

(Magnetic Resonance) The inversion time delay (TI) is the time period from an 180° inversion pulse to the excitation pulse. Sequences employing an inversion pulse are called inversion recovery pulse sequences. Since different tissues have different $T_1$, the inversion time affects the image contrast. Very often the inversion time is chosen to null a certain tissue or fluid for signal suppression, e.g. of fat using STIR or cerebrospinal fluid using FLAIR.

**Related Articles:** Fluid attenuated inversion recovery (FLAIR), Inversion recovery, Short tau inversion recovery (STIR)

**Related Articles:** Echo train length, Fast spin echo, Repetition time (TR), RF pulse, Spin echo, $T_2$-weighted


**Related Articles:** Echo train length, Fast spin echo, Repetition time (TR), RF pulse, Spin echo, $T_2$-weighted

**Related Articles:** Fluid attenuated inversion recovery (FLAIR), Inversion recovery, Short tau inversion recovery (STIR)

### FIGURE 1.44

(a) In an inversion recovery experiment, a 180° RF pulse inverts the longitudinal magnetisation. After a waiting time denoted $T_1$ (black, dark grey and light grey arrows) have undergone different amounts of longitudinal relaxation. This difference in potential signal is flipped into the transversal plane for readout, frequently by combining the 180° inversion pulse with a fast spin echo sequence. (b) Using phase-sensitive reconstruction the grey scale can be adjusted for negative as well as positive signal values, so that negative signal values will be dark, zero signal gray and positive signal values bright. (c) Another representation is provided by magnitude reconstruction, where absolute signal values are displayed.
Iodine

Iodine (General)

<table>
<thead>
<tr>
<th>Symbol</th>
<th>I</th>
</tr>
</thead>
<tbody>
<tr>
<td>Element category</td>
<td>Halogen</td>
</tr>
<tr>
<td>Mass number $A$ of stable isotope</td>
<td>127 (100%)</td>
</tr>
<tr>
<td>Atomic number $Z$</td>
<td>53</td>
</tr>
<tr>
<td>Atomic weight</td>
<td>126.90447</td>
</tr>
<tr>
<td>Electronic configuration</td>
<td>$1s^2 2s^2 2p^6 3s^2 3p^6 3d^{10} 4s^2 4p^6 4d^{10} 5s^2 5p^6$</td>
</tr>
<tr>
<td>Melting point</td>
<td>386.85 K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>457.4 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>$4.933 \times 10^4$ kg/m$^3$ (4.933 g/cm$^3$)</td>
</tr>
</tbody>
</table>

**History:** Although it is known that iodine was first isolated in 1811 by Bernard Courtois during the manufacture of saltpetre, there was some debate over who should be credited with identifying it as a new element. In December 1811, within days of each other, Sir Humphry Davy and Joseph Louis Gay–Lussac independently announced their discovery of the element.

**Isotopes of Iodine:** 37 isotopes of iodine exist. Only one, iodine-127, is stable and exists in nature as part of compounds such as iodides. Several unstable isotopes are used in nuclear medicine imaging and radionuclide therapies. These are as follows:

<table>
<thead>
<tr>
<th>Isotope of iodine</th>
<th>$^{123}$I</th>
</tr>
</thead>
<tbody>
<tr>
<td>Half-life</td>
<td>8.02 days</td>
</tr>
<tr>
<td>Peak decay energy</td>
<td>606 keV beta (positron); 364 keV gamma</td>
</tr>
</tbody>
</table>

**Clinical Applications:** Iodine-123 is used for imaging in nuclear medicine because of its nearly ideal photon energy for the scintillation camera, i.e. 159 keV (89%). The radionuclide also emits photons with higher energy resulting in a risk of septum penetration in the collimator.

**Related Articles:** Thallous iodide (Tl-201) and thallous chloride (Tl-203) are used in nuclear medicine for imaging of the heart, vascular system, and certain neuroendocrine tumours. Iodine-125 is used in low dose rate brachytherapy.

**Related Articles:** Thallous iodide (Tl-201) and thallous chloride (Tl-203) are used in nuclear medicine for imaging of the heart, vascular system, and certain neuroendocrine tumours. Iodine-125 is used in low dose rate brachytherapy.

**Medical Applications:** Nuclear Medicine Diagnostic Imaging – Both stable and radioactive iodine are readily taken up by cells in the thyroid gland. Radioactive isotopes of iodine with suitable emission properties and half-life are used as tracers for diagnostic imaging of the thyroid (and occasionally the kidney, although this is largely obsolete). The tracer, usually iodine-123 or iodine-125 in the form of sodium iodide, is injected intravenously and the thyroid gamma emissions are imaged with a gamma camera to assess thyroid function.

**Radionuclide Therapy – Iodine-131** is administered orally to patients with thyrotoxicosis or thyroid cancer, and taken up by the thyroid. The emitted beta radiation ablates thyroid tissue to reduce thyroid activity or remove cancerous cells. Iodine-131 is also used as part of Meta-Iodo-Benzyl-Guanidine (MIBG) therapy to ablate certain neuroendocrine tumours. Iodine-131 is used in low dose rate seeds for prostate brachytherapy.

Iodine-124

(Nuclear Medicine)

Element: iodine (halogen)
Isotopes: 55 < N < 88
Atomic number (Z): 53
Neutron number (N): 71
Symbol: 124I
Production: Cyclotron
Daughter: 124Te
Half-life: 4.176 days
Decay mode: EC (74.4%); β− (25.6%)
Radiation: β− radiation; γ-radiation followed by x-rays and electrons
Energy: hv = 511 keV (200%);
Positron range ~3 mm in tissue
Biological half-life: 80 days (thyroid), 8 h (other organs)
Critical organ: Thyroid (normal uptake 30%–35%)
\( \text{ALI}_{\text{mmax}} \) (50 mSv): 100 MBq
Absorbed dose (iodide): 4.5 mGy/MBq (thyroid)
Effective dose: 0.2 mSv/MBq (oral); 0.08 mSv/MBq (inhalation)

\[ \begin{align*}
\text{Iodine} & \quad 126.90447 \\
[\text{Kr}]d^{10}5^25p^5 & \quad 10.4513
\end{align*} \]

Clinical Applications: Iodine-124 is used for PET-imaging in diagnostic nuclear medicine. It is labelled to the same pharmacueticals as \(^{131}I\).

The clinical applications are the same as for \(^{123}I\)-MIBG for the detection of neuroblastoma and pheochromocytoma, \(^{124}I\) as iodide for imaging of thyroid tissue, and \(^{124}I\)-Ioflupane (DaTScan) for presynaptic neuro-imaging.

Related Articles: Iodine-123, Iodine-125, Iodine-131


Iodine-125

(Radiotherapy, Brachytherapy) Iodine-125 decays by electron capture to an excited state of tellurium-125 followed by decay to the ground state, giving 35.5 keV gamma rays and 27–32 keV x-rays. The half-life of iodine-125 is 59.4 days and the average energy 0.028 MeV. (Iodine-125 is produced by neutron activation of xenon-124, naturally occurring, resulting in xenon-125, which decays by electron capture to iodine-125, which in its turn decays by electron capture to tellurium-125.)

The most common use of iodine-125 sources is for permanent interstitial implants; it is also used for temporary applications with eye plaques. The low energy of iodine-125 minimises the radiation protection problems, and interstitial implants are performed using both manual and manual afterloading techniques. Note that iodine-125 is also used for in vitro nuclear medicine, e.g. in biological assays.

There are a number of commercially available designs of iodine-125 sources, often denoted seeds. The designs differ in the way the radioactive material is distributed within the encapsulation and the materials used, see Figure I.46. An important property of a seed is its visibility, as modern interstitial implant techniques are image-guided, using ultrasound, fluoroscopy, CT or MR. Thus, manufacturers use materials that show the seed in ultrasound imaging (surface properties), fluoroscopy and CT (x-ray attenuating properties) and MR imaging. The design of a seed, the choice of material and the distribution of the radioactivity all affect the dose distribution around the seed. Thus, even if the nominal seed strength is the same, seeds of different types and from different manufacturers will a priori have different dose distributions. Seeds are available with a variety of source strengths, and as loose seeds and strands, see Figure I.47.

For an update on commercially available seeds and their physical properties, accepted by the brachytherapy community as complying with AAPM recommendations, see the web site of the Radiological Physics Centre (RPC): [http://rpc.mdanderson.org/rpc]

Abbreviations: AAPM = American Association of Physicists in Medicine.

Related Articles: Brachytherapy sources, Interstitial brachytherapy, Permanent implant


Iodine-131

(Nuclear Medicine)

Element: iodine (halogen)
Isotopes: 55 < N < 88
Atomic number (Z): 53
Neutron number (N): 78
Symbol: 131I
Production: Fission product
Daughter: 131Xe
Half-life: 8.04 days
Decay mode: β−-decay
Radiation: β− radiation, followed by γ-radiation
Beta energy released: 970 keV (max)/323 keV (mean)
Gamma energy: 364.5 keV (81.5%) 80.2 keV (2.6%)
Absorption (HVL): 3 mm lead
Biological half-life: 80 days (thyroid), 8 h (other organs)
Critical organ: Thyroid (normal uptake 30%–35%)
\( \text{ALI}_{\text{mmax}} \) (50 mSv): 1 MBq
Absorbed dose (iodide): 530 mGy/MBq (thyroid)
Effective dose: 22 mSv/MBq (oral); 8 mSv/MBq (inhalation)

Clinical Applications: Iodine-131 has been used in nuclear medicine since the late 1930s, and is the standard radionuclide for thyroid treatment, as sodium radiodine. After oral administration of radiodine, the absorption from the GI-tract is rapid, 5% min−1, and nearly completed after 1–2 h. The rate of absorption may slow down if food is present, and it is influenced by the function of the thyroid. Iodine-131 is also used for, e.g. treatment of neuroblastoma and pheochromocytoma as \(^{131}I\)-MIBG (meta-iodobenzylguanidine) and occasionally as labelled monoclonal antibodies for treatment of specific cancers.
Ion collection

4.5 mm 4.7 mm

4.8 mm

Related Articles: Iodine-123, Iodine-124, Iodine-125


Ion

(Nuclear Medicine) An ion is an atom where the proton-electron ratio differs from one which gives it the property of either positive or negative charge. An ion with more electrons in the shell structure than protons in the nuclei has a negative charge and is referred to as anion. An ion with positive charge is called a cation (pronounced ‘cat-eye-on’).

Ion collection

(Nuclear Medicine) See Ion pair
ion pairs and free electrons may result in a recombination in which the electron is captured by the positive ion and returns it to a state of charge neutrality. Collisions between positive ions or free electrons which are created have a random thermal motion and consequently they diffuse away from regions of high intensity. In some collisions an electron may be transferred from a neutral gas molecule to a positive ion thereby reversing their role or an electron may be captured by a positive ion resulting in a recombination in which the electron is captured by the positive ion and returns it to a state of charge neutrality. In an ionisation chamber detector gas molecules are ionised by incident radiation and the observing ion pairs are collected using an electric field.


**Ion recombination**

(*Radiotherapy*) When ionising radiation interacts with a gas the positive ions or free electrons which are created have a random thermal motion and consequently they diffuse away from regions of high intensity having many types of collision. In some collisions an electron may be transferred from a neutral gas molecule to a positive ion thereby reversing their role or an electron may be captured by a positive ion returning to a state of charge neutrality. Collisions between positive ions and free electrons may result therefore in a recombination in which the electron is captured by the positive ion and returns it to a state of charge neutrality. Alternatively, the positive ion may undergo a collision with a negative ion and both ions are neutralised. The frequency of collisions is proportional to the product of the concentrations of the two species involved. The recombination rate can be expressed as

\[
\frac{dn}{dt} = -\alpha n^+ n^-
\]

where

- \( n^+ \) is the number density of positive species
- \( n^- \) is the number density of negative species
- \( \alpha \) is the recombination coefficient

Recombination is therefore significant only in regions in which the concentration of ions and electrons is high and this condition surely can be found along the track of an ionising particle where the ion concentrations are maximum before the ion diffusion. In the presence of an external electric field, as in the case of the sensitive volume of an ionisation chamber filled with air and exposed to radiations, the drift of the positive and negative charges represented by the ions and electrons constitutes an electric current and its intensity is influenced by the recombination process.

A recombination occurs when the positive and negative ions formed in the same track meet and recombine. This process is called initial or columnar recombination as the tracks tend to be linear in the gas. Initial recombination is a dose rate independent effect unless the ion density becomes so great that the field strength responsible for the ion collection is reduced. The initial recombination is small for the radiation dose rates encountered in the clinical use of ionising radiation. When ions from different tracks encounter each other on their way to the collecting electrodes and recombine the effect is called general recombination. This effect depends on how many ions are created per unit volume and per unit time and therefore it depends on dose rate since a greater density of ions of both signs increases the probability that they will recombine.

It is possible to distinguish experimentally between the initial and the general recombination by plotting the reciprocal of the observed ionisation current \( I \) against an suitable function of the collecting field strength \( X \).

For the initial recombination the function is given by

\[
\frac{1}{I} = \frac{1}{I_{sat}} + \frac{k}{X}
\]

while for general recombination the relation is

\[
\frac{1}{I} = \frac{1}{I_{sat}} + \frac{k'}{X^2}
\]

with \( k \) and \( k' \) constant.

The total recombination is determined by the sum of the initial and general recombinations.

In Figure I.48 the current voltage characteristics of an ionisation chamber are reported. No net current flows in absence of an applied voltage. As the voltage increases the electric field begins to separate the ions without avoiding the recombination effect. The combined effect of recombination and diffusion determine the shape of the rise to the saturation in the current voltage characteristic curve. The ion diffusion effect does not permit measurement of the saturation current and in practice the voltage \( V \) is raised to high values to minimise the losses.

One empirical relationship that permits the evaluation of losses due to the diffusion effect when the applied voltage is sufficiently high is given by

\[
\frac{1}{I} = \frac{1}{I_s} + k \frac{d^3}{V^2}
\]

where

- \( I_s \) is the saturated ionisation chamber
- \( d \) is the electrode separation distance in the chamber
- \( V \) is the applied voltage
- \( k \) is a constant characteristic of the specific chamber

**FIGURE I.48** Current–voltage characteristics of an ionisation chamber.
Ion therapy

Radiotherapy) Ion therapy is radiotherapy delivered with ions of light nuclei, often carbon, to treat deep seated tumours. The advantage over x-rays is the fact that the ions have a Bragg peak which means that the dose is delivered mostly over a small range of depth rather than with the classic exponential distribution of x-rays. Compared with protons, the energy deposition density is greater.

A magnetic scanning system may be used to direct the beam across the tumour in a raster scan, which coupled with energy modulation allows intensity-modulated radiotherapy (IMRT) with the ion beam. Energy modulation allows broadening of the depth of maximum dose in what is known as a spread out Bragg peak (SOBP).

In order for the ions to penetrate sufficiently deep to treat deep seated tumours, they must have energies of 300–400 MeV/nucleon typically. This is often achieved using large particle accelerators such as synchrotrons, meaning that this form of treatment is often delivered at large national facilities.

Ion therapy should not be confused with so-called negative ion therapy.

Abbreviations: IMRT = Intensity-modulated radio therapy and SOBP = Spread out bragg peak.

Related Articles: Proton therapy, Spread out Bragg peak, Heavy particle beams, Hadron therapy, Charged particle therapy

Ionisation

(Radiation Protection) The ejection of an electron from an atomic shell by the interaction of radiation. There are a variety of interactions that can result in ionisation. For example, x- or gamma-rays interact with matter via a number of different processes, of which two result in ionisation; Compton scattering and the photo-electric effect.

Ionisation presents a biological risk since the absorption of energy both from the incident radiation, and the kinetic energy of the ejected electron within cellular molecules can lead to malformation (as a result of single ionisations of the DNA molecule) or deterministic effects (as a result of gross cell killing from multiple ionisations within each cell).

Related Articles: Interaction, Photoelectric effect, Compton scatter

Ionisation chamber

(Radiation Protection) The ionisation chamber is the simplest gas-filled detector in which the electric current caused by the ions pairs created within the gas is measured. The X or gamma radiation passing within the gas interacts with its molecules through photoelectric, Compton or pair production process. The electrons accelerated by the electric field applied between electrodes (Figure I.50) produce the ion pairs consisting of positive ions and free electrons.

The intensity of the measured current depends on the external voltage value (Figure I.51). Its measurement is realised in the saturation region where all electric charge (ion pairs) is collected by the electrodes. The applied voltage is about 100 V and the ionisation current about $10^{-14}$ A. The guard rings are used to reduce the effects of the insulator leakage which would be added to the measured signal. Depending on the separation between the electrodes and the density of ion pairs in the gas some recombination of ion pairs in the gas may occur depending on the voltage.

The ion current is measured indirectly by using either sensing of the voltage drop (Figure I.52) across the resistor ($10^9$–$10^{12}$ Ω) or the voltage due to an electric charge gathered by the capacitor (Figure I.53). The integration method based on the measurement of current over a finite period of time can also be applied.

The air-equivalent chamber is used for the exposure rate $R_e$ estimation as the ratio of the measured saturated current intensity $I_s$ (A) to the mass $m$ (kg) of the air in the active volume of the detector:

$$ R_e = I_s/m \, (C/s/kg) $$

FIGURE I.49 Variation of the ionisation charge $Q'$ collected by an ionisation chamber versus applied voltage.

The dose deposited in the air inside the sensitive volume of an ionisation chamber is proportional to the charge $Q$ produced by radiations. Because of the recombination the charge $Q'$ collected by the biased electrode of the chamber is less than $Q$ and it depends on the collecting voltage. In Figure I.49 the variation of the ratio between $Q'$ and $Q$ is reported as a function of the collecting voltage. When the applied voltage increases the recombination effect decreases. $Q'$ = $Q$ at the saturation but it is not possible to increase indefinitely the applied voltage because of ion multiplication in the ionisation chamber gas and because of electrical breakdown of insulator used in the ionisation chamber construction. It is necessary therefore to determine a recombination factor to determine $Q$ at saturation.

Ionisation chamber

The scheme of an ionisation chamber: $E_1$ – cathode, $E_2$ – anode, $E_3$ – guard rings.
that are in thermal random motion. There are many types of collisions including charge transfer collisions. In the charge transfer collisions a positive ion can take an electron from the neutral molecule or a free electron can be attached to a neutral atom (molecule). During the collision between the positive ion and the free electron or a negative ion a neutral atom (molecule) may be formed. This process is called ion recombination. In such a situation the electric charge is lost and it will not contribute in the ionisation current measured by the detector.

There are two types of ionisation recombination loss:

1. Initial (columnar) recombination which occurs in a column of a track of an alpha particle or other strong ionising particles.
2. Volume recombination which occurs for ions and electrons moving towards the collecting electrodes. It increases with increasing irradiation. To minimise this effect it is necessary to apply a high electric field.

Related Article: Ion recombination


Ionising radiation

(General) Ionising radiation is any form of radiation, particulate or electromagnetic, which can cause ionisation, which occurs when an electron is emitted from an atom or molecule. The radiation must have sufficient energy to overcome the binding energy of the electron that is emitted. Ionisation leaves the atom or molecule in a charged state.

When the radiation does not have sufficient energy to cause ionisation then excitation will occur, this is when the radiation excites the motion of an atom or molecule or induces an electron to move from an occupied orbital state to a higher unoccupied state.

Particles (such as alpha particles and electrons) cause ionisation by electrostatic interactions. In tissue the average energy required to form an ion pair is about 34 eV.

Electromagnetic radiation can cause ionisation by one of three processes: Compton scattering, the photoelectric effect or by pair production. The energy required depends on the atom or molecule involved but generally electromagnetic radiation of more than a few electron volts is considered to be ionising radiation. This means that only photons in the middle to far ultraviolet region, x-ray or gamma ray regions can cause ionisation.

Abbreviations: eV = Electron volt.

Related Articles: Auger particles, Beta particles, Compton scattering, Internal conversion electrons, Photo-electric effect, Pair production

iPAT (integrated parallel acquisition technique)

(Magnetic Resonance) Integrated parallel acquisition technique (iPAT) is a vendor term (Siemens) for partial parallel imaging (PPI) techniques, where determination of the coils’ sensitivity profiles is integrated into the scan. The information for the so-called auto-calibration is obtained from a specified number of central k-space lines representing a low-pass filtered signal. Since these additional signals are eventually employed in the reconstructed image (e.g. in the GRAPPA algorithm), auto-calibration implies a reduced loss of signal-to-noise-ratio (SNR).

Related Articles: Partial parallel imaging, GRAPPA
**Iridium-192**

*(Radiotherapy, Brachytherapy)*  Iridium-192 is a reactor produced radionuclide (neutron activation of stable Iridium-191) with a half-life of 73.83 days. Iridium-192 decays via electron capture and beta emission to excited states of osmium-192 and platinum-192, which decay to stable states by gamma emission (x-rays are also produced). The photon spectrum of iridium-192 is complex, containing a large number of energy levels, with a maximum of more than 1 MeV. Iridium-192 is a photon source and the average photon energy is often given as 0.38 MeV for an encapsulated high dose rate source (the exposure weighted effective energy is 397 keV). The decay beta radiation is mostly absorbed in the encapsulation. It is to be noted, that secondary photons and electrons are produced in photon interactions both in the source material and the encapsulation; care must be taken in source strength measurements.

**Types of Iridium-192 Sources:** Iridium-192 is widely used for temporary implants, and there are sources available for most types of brachytherapy; intracavitary and interstitial, high dose rate, low dose rate and pulsed dose rate techniques. Iridium is not used for permanent implants due to the rather high averages photon energy causing radiation protection problems (compared to the low photon energy of iodine-125)

Low dose rate sources: Iridium sources are available as platinum coated wires, and as small stranded seeds. These sources are generally handled with afterloading techniques, both manually and remotely controlled. (The Paris system for interstitial implants was designed for low dose rate iridium wires.)

High dose rate and pulsed dose rate sources: Iridium-192 is well suited for high dose rate sources due to its high specific source strength. Small high intensity sources make it possible to used small applicators together with short treatment times. Typical source strength for a high dose rate source is 40.70 mCi (1509 GBq = 10 Ci apparent activity), and sources with sizes in the order of 0.7 mm and length 3.5 mm, encapsulation 0.9 mm and length 4.5 mm, are available, see the drawing in Figure I.54.

High dose rate and pulsed dose rate sources can only be used with remote afterloading units. The high dose rate source in Figure I.54 (GammaMed Plus, Varian) is welded to a drive cable, and source stop positions and dwell times are controlled by the precision drive motor in the afterloading unit and the accompanying treatment control software. Figure I.55 shows the position of the source at its first, most distal, stop position in an applicator, together with the marker wire mimicking stop positions with a step size of 1 cm.

Due to the relatively short half-life of iridium-192, the source is usually replaced four times per year, to keep treatment times short. Each source exchange requires extra quality control tests besides the ordinary daily tests, and determination of source strength and determination of source stop positions are the two most important parts of the specific source exchange QA programme.

**Related Articles:** Brachytherapy sources, Remote afterloading units

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**Iris diaphragm**

*(Diagnostic Radiology)* The iris diaphragm is normally placed in front of the TV camera of a digital fluoroscopic system (such as DSA). The device stays between the image intensifier output and the TV camera (it is similar to the aperture of a normal photographic camera). The iris diaphragm provides exact light beam size, hence overall light intensity, to the TV camera input/target.

The adjustment of the iris diaphragm is related to the proper functioning of the TV camera tube (such as Vidicon type), avoiding overexposure (saturation) or underexposure of its target. This way the TV camera will always function within the linear part of its characteristic.

---

**FIGURE I.54** High dose rate source for the GammaMed Plus afterloading unit (Varian).

**FIGURE I.55** The first, most distal, stop position of this source in an applicator. Also shown is a marker wire inserted in the applicator, indicating source stop positions with a step length of 1 cm.

The Iris diaphragm influences also the signal to noise ratio (SNR) of the final image. If the aperture of the Iris diaphragm is smaller less light from the Image Intensifier will reach the TV camera. This could drive the automatic feedback system (automatic brightness control) to increase the radiation dose, hence the overall light intensity. This will increase the patient dose but at the same time will decrease the quantum noise and produce image with better SNR.
**Related Articles:** Automatic brightness control, Digital subtraction angiography (DSA), x-ray television, Vidicon TV camera

**IRPA**  
(General) See International Radiation Protection Association

**Irradiated volume**  
(Radiotherapy) The irradiated volume in radiotherapy is the tissue volume that receives a dose that is considered significant in relation to normal tissue tolerance. The significance level will depend on the normal tissue type. For more information about significant dose please see the articles on Parallel organ, Serial organ, and Tolerance.

The use of ‘irradiated volume’ as a planning volume was proposed by the ICRU in Report 50 (with addendum 62). This report provided a common framework on prescribing, recording and reporting therapies, with the aim to improve consistency and inter-site comparability. It details the minimum set of data required to be able to adequately assess treatments without having to return to the original centre for extra information (Figure I.56).

**Related Articles:** Planning target volume (PTV), Clinical target volume (CTV), Treated volume, Gross tumour volume (GTV), Parallel organ, Serial organ, Tolerance


**Irregular field**  
(Radiotherapy) This is a term generally given to an open field that has been shielded using low melting point alloy, lead or multileaf collimators (MLC) (Figure I.57).

**Related Articles:** Multileaf collimator, Equivalent square

**ISIS**  
(Magnetic Resonance) See Image selected in vivo spectroscopy (ISIS)

**Isobars**  
(General) Isobars are nuclides with the same mass number; that is the same number of nucleons (protons and neutrons). The nuclides will therefore be of different elements.

At lower mass numbers normally only one of the isobars is stable – an example, N-14 is stable but all the other isobars of N-14, e.g. C-14, are radioactive. However, with increasing mass there are more examples of several isobars being stable – an example: Ar-40 and Ca-40 are all stable. K-40 is not stable but has a very long half-life.

**Related Articles:** Isotope, Isotone, Nucleons

**Isocentre**  
(Magnetic Resonance) The point at the centre of the magnet of an MR scanner is called the isocentre. It defines the origin $(x, y, z = 0,0,0)$ of a right handed coordinate system (Figure I.58). This coordinate system can be used to describe the directions of the gradient fields (see related article) as well as many other parameters related to pulse sequences, the position of the patient and geometrical properties of the MR images. Also, the isocentre is located at the centre of a usually spherical or elliptical volume with a very homogeneous $B_0$ field. This volume is where the imaged part of the patient is located during an MR examination. $B_0$ as well as the resonance frequency of the MR scanner are defined only at the isocentre. Thus, the isocentre is not a point in space but represents the maximum useable field-of-view (FOV).

**Related Articles:** Gradient coils, Gradient fields, Field-of-view

**Isocentric technique**  
(Radiotherapy) This is the ‘point’ about which all the various axes of rotation of the treatment machine meet. The gantry of the accelerator, the collimators, and the couch all rotate about this point. The room lasers are also aligned to meet at this position. There are also two types of isocentre – mechanical and radiation. The mechanical isocentre relates to the point about which all the equipment rotates about, while the radiation isocentre is the point about which beams delivered from different gantry angles meet. Ideally these points should be the same.

It is becoming increasingly common for treatments now to be delivered isocentrically – that is with the centre of the tumour positioned at the isocentre, rather than with a fixed SSD.

**Related Articles:** Mechanical isocentre, Radiation isocentre, Isocentric technique

**FIGURE I.56** Definition of target volumes as in ICRU 50.

**FIGURE I.57** Regular and irregular fields.

**FIGURE I.58** The coordinate system of a cylindrical magnet.
Isocentric technique
(Radiotherapy) This technique typically places the centre of the target at the isocentre position for all treatment fields, and so the source to skin distance (SSD) will change for different beams depending on the depth of tissue the beam must penetrate. Another name for this technique is the fixed source axis distance (SAD) technique.

The alternative technique to this is to maintain a constant SSD to the skin surface and so the target will be at varying distances for each field.

Abbreviations: SAD = Source axis distance and SSD = Source to skin distance.

Related Articles: Isocentre, Multiple isocentre treatment, Source axis distance (SAD), Source-to-skin distance (SSD)

Isodose curve
(Radiotherapy) An isodose curve is a line of constant absorbed dose. The isodose curves are generally drawn at regular intervals of absorbed dose and are expressed as percentage of the dose at a normalisation point (i.e. 80%, 70%, etc.). A set of isodose curves is called isodose chart. These refer commonly to principal planes that are planes which contain the beam axis. Measurements of the isodose curves should be made in a water tank large enough to permit a full scatter condition to the point at which measurements are being made or in a tissue equivalent phantom. In Figure I.59 isodose curves are shown for a normal incidence of the photon beam on a homogeneous phantom. The isodose curves could be measured by an ionisation chamber, a solid-state dosimeter, thermoluminescence dosimeters and films. A precision carriage could be mounted within the water tank allowing the 3D placement of the detector at any position in the tank under the remote control of a computer.

Isodose rate surface
(Radiotherapy, Brachytherapy) Volumes in Brachytherapy: An isodose surface is a surface that defines a volume enclosed by that surface.

Similarly an isodose rate surface is a surface that defines a volume enclosed by that surface.

Isodose and isodose rate surfaces are used in both permanent and temporal implants at low dose rate. For high dose rate and pulsed dose rate implants, only isodose surfaces as used.

Examples: the reference volume of ICRU Report 38 is the volume enclosed by the reference isodose surface. ICRU Report 58 defines the reference volume as the volume encompassed by an isodose surface in relation to the mean central dose.

Abbreviation: ICRU = International Commission on Radiation Units and Measurements.

Related Articles: Reference isodose, Reference volume, Paris system, Manchester system


Isodose shift method
(Radiotherapy) The isodose shift method is a simple method to correct for oblique beam incidence or non-uniform patient contours across the beam. In the oblique entry situation, the dose distribution will be different to that as measured for a flat phantom surface with a beam incident at the normal angle to the surface.

In this method the value of the dose at a point P is shifted on a vertical line by a quantity \( h \times k \), where \( k \) is a factor to take account of the energy of the beam used and \( h \) is a measure of the thickness of tissue either missing or in excess at that point relative to the central axis (Figure I.60).

In situations where there is missing tissue, the quantity \( h \) will be a positive value and so the isodose lines are shifted deeper into the patient or phantom, while the opposite is true for excess tissue where \( h \) has a negative value.

The factor \( k \) is always less than 1 and is 0.7 for cobalt beams and energies less than 5 MV, 0.6 for energies between 5 and 15 MV, 0.5 for energies between 15 and 30 MV, and 0.4 for energies greater than 30 MV.

Related Articles: Oblique incidence, Obliquity, Obliquity effect

Isodose surface

(Radiotherapy) Isodose surfaces are a family of surfaces in 3D which connect all points of equal dose in an irradiated volume as shown in Figure I.61. A 2D plane taken through these surfaces produces a family of isodose curves or isodose lines as shown for a lung patient in Figure I.62.

Isoeffect

(Radiation Protection) The isoeffect approach can be used to study combined effects of mixed radiations. In principles it is the same method used to study the effect of drugs and antibiotics. The method consists of in vitro and in vivo measurements. For beams containing radiation components belonging to a range of linear energy transfer (LET) it is important to study the interaction of these radiation components of different qualities (crucial in the planning of the radiotherapy treatment with neutron and charged particles). Studies have demonstrated that, in certain cases, the combined effects of high and low LET are greater than what could be calculated considering the independent effects of each component. There are models to evaluate the synergism of radiations of different LET. There are also showing that the simple isoeffect relation is applicable to a certain class of radiobiological data. A detailed investigation is required for each particular combination of radiations.

Hyperlink: https://www.IAEA.org; Pub med and med line

Isomeric transition (IT)

(General) An isomeric transition is the transition of a nucleus in an excited metastable state to the ground state with the emission of a gamma ray.

A nucleus in an excited metastable state is called a nuclear isomer. After radioactive decay of many radionuclides, the daughter nuclide is in an exited nuclear state. In most cases the nucleus then de-excites virtually instantaneously, in less than a picosecond, with the emission of gamma radiation and the end result is the daughter in its ground state.

However, in a small number of radionuclides the excited daughter has a relatively long lifetime – measured in seconds, minutes or hours. These long-lived excited states are called ‘metastable levels’ and are designated by the use of the suffix or superscript m (for metastable). An example is the long-lived excited state of technetium-99, which is designated technetium-99m ($^{99m}$Tc). Particulate radiation is only emitted in the form of electrons as a result of internal conversion of the gamma radiation.

Technetium-99m is widely used in nuclear medicine and is the excited daughter of the radionuclide of molybdenum – see Figures I.63 and I.64.

Related Articles: Internal conversion, Gamma radiation, Radioactive decay, Radionuclide
Isomers

(Isomorphic) In physics, an isomer is each of two or more atomic nuclei with the same atomic number and mass number but different energy states. Derived from the Greek *isomerês* – ‘sharing equally’. The term nuclear isomer is often used to distinguish it from other types of isomers.

A particular type of nuclear isomer is when a nucleus is in an excited metastable state which has a relatively long lifetime – measured in seconds, minutes or hours. An isomeric transition is the transition of a nucleus in an excited metastable state to the ground state with the emission of a gamma ray.

**Related Articles:** Isomeric transition, Radioactive decay

Isotopes

(General) Nuclides are isotopes if they have the same number of neutrons in the nucleus. The nuclides may be stable or radioactive.

Helium (\(^{4}\text{He}\)) has two protons and two neutrons but deuterium (\(^{2}\text{H}\)) also has two neutrons and so these two nuclides are isotopes.

The term isotope is derived from the term isotope (nuclides having same number of protons) by replacing the p with an n.

**Related Articles:** Isotope, Nuclide

Isotope

(General) An isotope is one of a group of nuclides that have the same proton (atomic number) mass. This group is composed of different nuclides of the same element that differ in neutron number and, in some cases, nuclear state. Because they differ only in their complement of neutrons they have similar physical properties and the same chemical properties. Isotopes are either stable or unstable (radioactive) in which case they may be referred to as radioisotopes. Radioisotopes undergo radioactive decay with a specific half-life. In general, the greater the difference in mass between an unstable and a stable isotope, the greater is the instability of the radioactive isotope and the shorter the half-life.

**Related Articles:** Radioisotope, Radiouclide, Half-life, Radioactive decay

Isotopic

(General) For isotropic radiation the intensity is uniform in every direction, i.e. not dependent on the direction. For example, the particles and photons emitted during decay are emitted isotropically. The antonym for isotropy is anisotropy, i.e. the radiation intensity is direction dependent.

Isotopic emission

(Nuclear Medicine) The intensity from a radioactive source with isotropic emission is independent of direction, i.e. the source has no preferred direction of radiation. Isotopic emission is one of the fundamental physical rules and applies to all kinds of radioactive decay. Even though the emission from a single radioactive nucleus or small sample is isotropic the emission from a radioactive sample seldom is because of sample spread and self-absorption within the sample.

Isowatt circuit

(Diagnostic Radiology) See Automatic brightness control (ABC)

Iterative algorithm

(Nuclear Medicine) See Iterative reconstruction methods

Iterative reconstruction methods

(Nuclear Medicine) The iterative reconstruction method is an alternative to filtered backprojection (FB), but has until recently been discarded as too computationally intensive. But as computer speed has increased the method is becoming a more common alternative.

The reconstruction starts with an estimation of the true image. This estimation is often a simple image, such as a blank or uniform image. The algorithm will eventually approach the truth by performing gradual approximations, or iterations. The next step is to acquire the projection that would have been measured to attain the estimated image. This process is an inverse backprojection, or forward projection and it is basically a summation along the potential ray paths for each projection. The projections from the estimated image are then compared to the measured projections. The two sets of projections are not likely to agree at an early stage since the true image differs from the initial estimation image which is a blank or uniform image. But the difference between the two projection sets can be used to update the estimation image. This update process is repeated until the similarity between the two images reaches a certain threshold.

As described earlier, the concept of iterative reconstruction algorithms consists of two basic components: (1) the method for comparing the true projections and the estimated and (2) the method where the estimation is updated according to the comparison. Using iterative reconstruction methods it is possible to give different weights to different data in the projection. Specifically, give higher weight to high count elements and less weight to low count elements to increase the signal-to-noise ratio. Other advantages when using iterative reconstruction methods is that it is possible to model physical properties, like scatter events and septal penetration.

**MLEM (Maximum-Likelihood Expectation-Maximisation):**

This algorithm will produce the ‘most likely’ (maximum likelihood) source distribution derived from the observed projections. In this reconstruction method all projection angles are used to update the estimated image. This procedure slows down the process and it requires several iterations before a reasonable image is acquired. The time consumed in the backprojection during each iteration is proportional to the number of projections used. OSEM is a commonly-used algorithm used that has found a way around this problem.

**OSEM (Ordered Subsets Expectation-Maximisation):**

Instead of using all the projections to estimate an image, the OSEM algorithm only uses a subset of projections, meaning four orthogonal projections. The projections are used to calculate a new estimated image. The procedure is repeated with four new projections to make a new estimated image. This algorithm is faster and produces images with high spatial resolution after one iteration, unlike MLEM where the image resolution is enhanced for every iteration using all projections.

To get a more elaborate explanation of iterative image reconstruction methods the reader is referred to respective chapters in the following two references.


**Related Articles:** Filtered backprojection, Signal-to-noise ratio

IUPESM

(General) See International Union for Physical and Engineering Sciences in Medicine
**J coupling**  
*(Magnetic Resonance)* An alternative term for scalar magnetic coupling, in which *J* is the coupling constant characterising the interaction between nuclei.  
*Related Article:* Magnetic coupling

**Joule**  
*(General)* The joule (symbol J) is the unit of energy (work, heat) in the International System of Units (SI), equal to the product of the force in newtons (SI unit – symbol N) and the distance in metres (SI unit – symbol m).  

The Joule is therefore a derived SI unit that can more properly be defined in terms of base SI units, expressed as  

\[ 1 \text{ J} = 1 \text{ kg} \times (\text{m}^2/\text{s}^2) \]

where  
- kg is kilogram (mass)  
- m is metre (distance)  
- s is second (time)

The Joule is named after the English physicist James Prescott Joule (1818–1889).  
*Related Article:* SI (Système International)  

**Justification**  
*(Radiation Protection)* The first principle of protection against ionising radiation for workers, patients and members of the public, specified by the International Commission on Radiological Protection (ICRP), is justification.  

Any practice involving the exposure of staff, patients or the public to ionising radiation must be justified – a process that evaluates the benefits to the exposed individuals or to society that result from such a practice against the risks associated with the potential adverse radiation effects of the exposure. The most recent ICRP Recommendations (ICRP 103, 2007) has the following definition: ‘The Principle of Justification: Any decision that alters the radiation exposure situation should do more good than harm’.  

Once a practice has been justified, then the second and third principles must be applied – i.e. Optimisation and Dose limitation.  
*Related Articles:* Adverse radiation effects, Dose limitation, International Commission for Radiological Protection (ICRP), Optimisation  
K absorption edges

(General) When the energy of an incoming photon exactly matches that required to eject a K-shell electron, strong resonant photoelectric absorption occurs, producing what is known as an absorption edge.

Wavelengths shorter than the K absorption edge have sufficient energy to eject K orbital electrons; wavelengths slightly longer than the absorption edge do not. As photon energy is increased beyond the absorption edge, the interaction cross section decreases.

K absorption edges have a number of important applications in diagnostic radiology.

The energy at which the K absorption occurs is determined solely by the atomic number of the absorber. This means that sequences of materials with adjacent atomic numbers can be used to filter photon beams, permitting transmission of only photons within a narrow band of energies. Following the absorption of photons at the K-edge, characteristic x-rays are emitted. These will not be absorbed by the emitting material (their energy is slightly below the K-edge), but can be absorbed by materials with slightly lower atomic numbers. This principle is used to ‘harden’ beams generated by x-ray tubes, using combinations of copper, aluminium and bakelite filters.

Absorption edges also provide the fundamental mechanism behind x-ray contrast media. Radio-opaque contrast media such as iodine and barium have K absorption edges matched to the spectrum of x-rays leaving the patient.

Related Articles: Absorption cross section, Absorption by contrast media, Filter, Photoelectric absorption

K-edge

(General) See K absorption edges

K-edge metal filter

(Diagnostic Radiology) The K-edge filter is used to attenuate the high photon energy range of an x-ray spectrum. It is based on the principle that a material produces maximum x-ray attenuation by the photoelectric process for photon energies just above the binding energy of the K-shell electrons.

The energy of the K-edge (K-shell electron binding energy) is determined by the atomic number (Z) of the material. Therefore, a filter material can be selected to adjust the K-edge attenuation to the desired photon energy as illustrated in Figure K.1.

Molybdenum (Z = 42) is the most common K-edge filter material and is used in mammography to attenuate the spectrum above its K-edge energy of 20 keV. Rhodium (Z = 45) is an alternative filter material with a K-edge energy of 23.22 keV. Switching from the molybdenum to the rhodium filter extends the x-ray spectrum, making it more penetrating and more useful in the imaging of dense breast.

Related Articles: K-edge, Filtration total, X-ray beam filtration

Hyperlink: http://www.sprawls.org/resources/MAMMO

FIGURE K.1 A K-edge filter that attenuates the x-ray photon energies above the K-edge energy. (Graphs courtesy of Sprawls Foundation, www.sprawls.org)

Kerma

(Radiation Protection) Kerma (kinetic energy released per unit mass) is used to describe energy loss in a medium. Thus, air kerma represents the kinetic energy transferred to charged particles per unit mass of irradiated air when indirectly ionising (uncharged) radiations such as photons traverse the volume of air.

Kerma is the starting point for determining the energy deposition by a given type of radiation in an absorbing medium and varies according to radiation type and physical characteristics of the medium itself.

Kerma is formally defined as the kinetic energy released per unit mass:

$$K = \frac{dE}{dm}$$

where $dE$ is the sum of the kinetic energies of all the charged particles liberated by uncharged ionising radiation in a mass $dm$. This includes the energy of any Auger electrons.

The units of kerma or air kerma are J/kg or Gray (Gy).

As an approximation, the absorbed dose can be related to the kerma taking into account the fraction of the energy released in the mass, which is not absorbed within that mass:

$$D_{abs} = K (1 - g)$$

where $g$ is the fraction of energy of the charged particles liberated by the photons, which is lost in radiative processes – i.e. where particles are not fully stopped within the mass. At lower energies, $g$ is relatively small, implying that transient charged-particle equilibrium has been quickly reached, whereby the energy of charged particles
The convolution operation is usually denoted by ∗: in which the primary photon interactions are dealt with separately approach involves the convolution–superposition class of methods, the dose distribution is modelled from first principles. One such within the medium may be described by the graph in Figure K.2. Related Articles: Absorbed dose, Transient charged-particle equilibrium, Energy deposition Kernel (General) Kernel is an image processing term associated with filtering. In order to apply a filter to an image, a kernel matrix containing multiplication factors) is applied to every pixel in the image. Each pixel and its neighbours are multiplied by these factors and then the pixel is replaced by the sum of the products. This operation is mathematically known as convolution. It is defined as follows:

\[ g(x) = \int_{-\infty}^{\infty} I(x')h(x - x')dx' \]

where
- \( I(x) \) is the input image
- \( g(x) \) is the result
- \( h(x) \) is the convolution kernel

The convolution operation is usually denoted by ∗:

\[ g(x) = I(x) ∗ h(x) = h(x) ∗ I(x) \]

Equation K.1 shows the superposition equation for general calculation of dose at a point \( r \).

In Equation K.1, the product of the mass attenuation coefficient and the primary energy fluence, denoted by \( A \), is the TERMA (total energy released per unit mass), which represents the dose to the irradiated medium by the interaction of primary photons, and which can be readily calculated by ray tracing through the medium in a similar way to determining the radiological depth.

The term \( B \) is referred to as the kernel and describes the pattern of spread of energy away from the interaction point. Each point in the kernel describes the dose deposited, by scattered photons and electrons set in motion, as a fraction of the TERMA at the interaction point. These kernels can be obtained from deconvolution of dose distributions or by direct measurement, but the most common approach is to generate the kernels using Monte Carlo techniques (Mackie et al., 1988). The kernels can take the form of point, pencil or planar kernels and can be further decomposed into contributions from primary dose, single scatter dose and multiple scatter dose.

In the special case of spatially invariant kernels, Equation K.1 simplifies to a convolution integral where \( K(r - r') \) replaces \( K(r, r') \). The integral can then be efficiently evaluated using Fast Fourier Transforms, at a cost of reduced accuracy compared to the superposition integral.

The convolution–superposition class of dose calculations have been widely implemented into commercial treatment planning systems using point kernels, pencil beam kernels, dose deposition kernels and collapsed cone approximations (Figure K.3). Heterogeneities are accounted for by scaling the kernels for different tissues. Polyenergetic beams are accounted for by using a weighted sum of monoenergetic kernels.


Keyhole imaging (Magnetic Resonance) Keyhole imaging is a technique to speed up the acquisition of dynamic sequences in MRI. Normally, the entire k-space for each image is collected in a dynamic sequence. However, much k-space information in each subsequent image is identical in dynamic sequences; for example the information about the vessel edges in an angiography imaging sequence is likely to remain the same throughout the imaging time, if the vessel is not moving.

This is utilised in the keyhole technique. Instead of collecting all k-space phase encoding lines in each image, (morphological) information from a reference image is used instead.

The central part of k-space holds most of the contrast information in an image. In contrast, the outer parts of k-space depict the distinct borders in the image, such as vessel walls. When imaging, for example contrast agent going through vessels, a complete image
is acquired in the beginning before the contrast injection. During the contrast injection, only the central lines of $k$-space in the phase-encoding direction are collected. The missing outer parts of $k$-space are filled with data from the reference image (Figure K.4). In this way, information about the vessel signal (the contrast) is collected and updated during the examination, while information about vessel borders is taken from the reference image.

The keyhole technique is often used to increase the temporal resolution when imaging dynamic processes. However, organ motion should be avoided in order not to compromise spatial localisation.

**Related Articles:** Angiogram, $k$-space, Phase encoding


**Kit**

(*Nuclear Medicine*) Kit, or cold kit, is a prefabricated product containing the nonradioactive chemicals needed to produce a specific $^{99m}$Tc-radiopharmaceutical after adding the required activity $^{99m}$Tc-pertechnetate. The chemical substances in the sterile vial (kit) consists of the complexing agent (ligand), reducing agent, for example stannous chloride, stannous fluoride, or stannous tartrate, stabilisers, dispersing agents, transfer ligands and buffers, all in an nitrogen atmosphere. The solution is lyophilised to remove all water to extend the stability of the kit. Pertechnetate obtained from the generator does not label to the pharmaceutical without reduction to a lower valency state by the reducing agent. The stannous ion, however, is easily oxidised and therefore the kit is preserved in a nitrogen atmosphere.

The preparation of the radiopharmaceutical is performed by the injection of a sterile solution of $^{99m}$Tc-pertechnetate to the kit vial by using aseptic techniques. A breather needle should not be used because oxygen may affect the labelling procedure. Excess pressure can be avoided by withdrawing an equal volume of gas with the same syringe. The labelling instructions for Myoview™ (GE Healthcare) however state the use of a breather needle since nitrogen can influence the radiochemical purity. Some radiopharmaceuticals should also be heated in a boiling water bath in the labelling procedure. The instructions of the labelling procedure given by the manufacturer should be followed strictly.

Klein–Nishina differential cross section

(Radiation Protection) The Klein–Nishina differential cross section attempts to describe mathematically the result of interactions between incident photons of ionising radiation with single electrons in the absorbing medium. It describes the probability that a photon will be scattered to any particular angle, with associated transfer of energy to the electron.

The probability of scatter to any particular angle, for a given energy of incident photon, can be represented graphically thus (Figure K.5).

The Klein–Nishina differential cross section and equation, Bethe–Bloch equation, together with the, Molière scattering theory and others, attempts to describe interactions between ionising radiation and matter at an atomic level, and forms the mathematical basis for radiation dosimetry based on Monte Carlo statistical modelling.

Related Articles: Klein–Nishina equation, Bethe–Bloch equation, Molière scattering theory

Klein–Nishina equation

(Radiation Protection) The Klein–Nishina equation describes mathematically the probabilities of interactions occurring when photonic ionising radiation is incident on an absorbing medium.

For more detail, see the article on Klein–Nishina differential cross section.

Related Article: Klein–Nishina differential cross section


KLM equivalent circuit model

(Ultrasound) The Krimholtz, Leedom, Matthaei (KLM) model is an equivalent circuit model that describes the electro-acoustic coupling for a piezoelectric transducer. It is a development of the Mason model, mainly by the separation of the acoustical and electrical parts of the transmission process. In Figure K.6, the electrical part is in the lower part of the figure (port 3) and the acoustical part in the upper part (ports 1 and 2). Port 1 represents the transmission into water or the body, while port 2 represents the acoustic backing. The load applied here can be used to modify the bandwidth and sensitivity. If desired, additional loads can be added on port 1 representing matching layers, etc.

In the lower right part of the figure is shown how the KLM model can be collapsed into a single ABCD matrix between the electrical port and the forward acoustic load. Such a matrix can easily be implemented, for instance, in MATLAB®.

Klystron

(Radiotherapy) Klystrons are used in linear accelerators as sources of high-power microwaves, which accelerate the electrons in the waveguide. An alternative option for the microwave source is a magnetron.

A klystron is essentially a microwave amplifier that requires an initial source of low-power microwaves, and is illustrated in Figure K.7. An electron cloud, produced by a cathode filament, accelerates towards the zero potential of an electron beam collector via a series of buncher cavities. The first cavity receives low-power microwaves such that the electric field varies sinusoidally across the cavity walls. This causes the electrons to bunch together as they undergo varying acceleration, and move at a frequency depending on the resonant frequency of the cavity. The electrons then travel through connecting, magnetically steered drift tubes towards the final catcher cavity, which will resonate at the electron arrival frequency, and produce the amplified RF wave. This field can then be transferred to the main accelerating waveguide via intermediate waveguides.

Klystrons are relatively large compared to magnetrons and hence normally are mounted behind the gantry. This complicates the transfer of the microwaves from the klystron to the waveguide, which is mounted within the gantry. They also create a source of kilovoltage x-rays from the surroundings, absorbing the accelerating electrons. However, they are more reliable and hence need replacing less often, which offsets the greater initial cost. They are also more stable, and do not suffer from sensitivity to the Earth’s magnetic field fluctuations like magnetrons do.

Related Articles: Linac, Magnetron

FIGURE K.5 (See colour insert.) Distribution of scattered photons at various energies.

FIGURE K.6 The KLM equivalent circuit model. In the upper part of figure are the two acoustic ports 1 and 2, connected to the electrical port 3, lower left. To the lower right is an ABCD matrix representation of the model, between port 3 and port 1.

FIGURE K.7 A typical klystron arrangement.
**k-space**

*(Magnetic Resonance)* The *k*-space is a formalism to describe gradient encoding of the MRI signal prior to Fourier transform (independently introduced in 1983 by Ljunggren [1983] and Twieg [1983]).

The position in *k*-space of each complex data point time is given by the integral of gradients applied between time of excitation and acquisition:

\[
\vec{k} = \gamma \int dt \vec{G}(t)
\]

where \(\gamma\) is the gyromagnetic ratio. The *k*-space of a slice is two-dimensional (2D); three dimensions encode a volume. The MR image is calculated by a 2D or 3D discrete Fourier transform:

\[
\text{DFT} \quad S(\vec{x}) \Leftrightarrow \hat{S}(\vec{k})
\]

Thus, the spatial distribution of signal in real space \(S(\vec{x})\) corresponds to a distribution in *k*-space \(\hat{S}(\vec{k})\). \(\hat{S}(\vec{k})\) is usually a complex number (comprising magnitude and phase).

In analogy to optical imaging, the ‘spatial frequencies’ are called *k*-vectors and correspond to plane-wave modulations along their direction. Most of the image signal is located around the *k*-space origin. High frequencies encode fine detail in the image. Thus, the pixel resolution is inversely related to *k*-space coverage; the field-of-view (FOV) to *k*-space resolution. In standard spin-warp phase encoding, *k*-space and image space have the same number of rows and columns, unless zero-filling, regridding, partial acquisition is applied. The *k*-space of a magnitude image (after zeroing the phase by the modulo-operation) has hermitean symmetry.


**k-space trajectories**

*(Magnetic Resonance)* The *k*-space trajectory is the path traced through *k*-space in a given MRI imaging sequence. Any given point in 2D *k*-space has coordinates \((k_x,k_y)\), representing spatial frequencies in the \(x\) and \(y\) directions, respectively. Being ‘at a point in *k*-space’ means that, at a particular instant, spin phase angles across a slice in an object have a spatial distribution represented by the particular spatial frequencies \((k_x,k_y)\) at that point in *k*-space. A signal sample taken at that instant and written into the location in *k*-space is related to the amplitude of that spatial frequency component \((k_x,k_y)\) in an image of that slice. By sweeping through *k*-space and filling it with signal samples, sufficient data is acquired such that a Fourier transform of *k*-space will recover an image of a slice through the object.

The gradient fields control the *k*-space trajectory in the \((k_x,k_y)\) plane. Representing location in *k*-space as a time-varying vector \(\vec{k}(t)\), and the gradient field as a time-varying vector \(\vec{G}(t)\), the trajectory is given by

\[
\vec{k}(t) = \frac{\gamma}{2\pi} \int_0^t \vec{G}(t) \cdot dt
\]

where \(\gamma\) is the gyromagnetic ratio. Separating the \(k_x\) and \(k_y\) components of the vector \(\vec{k}(t)\) along the \(x\) and \(y\) directions in *k*-space, and defining one gradient direction as phase encode and the orthogonal direction as frequency encode, the trajectory can be rewritten in simplified form as

\[
k_x(t) = \frac{\gamma}{2\pi} \int_0^t G_{\text{freq}}(t) dt
\]

\[
k_y(t) = \frac{\gamma}{2\pi} \int_0^t G_{\text{phase}}(t) dt
\]

A Cartesian trajectory is illustrated in Figure K.8, using the example of a spin-echo sequence. The time course of the trajectories as described by (K.3) and (K.4) in the preceding text can be clearly seen, with the phase encode gradient controlling row to row movements in *k*-space and the frequency encode gradient sweeping the trajectory across columns in *k*-space.

![Figure K.8](image-url) Cartesian *k*-space trajectory in a spin-echo sequence. Simultaneous applications of the phase encode and frequency encode gradients sweeps the trajectory from A to B. The 180° pulse flips the trajectory to point C at the start of a row, and the second application of the frequency encode gradient sweeps the trajectory along a row from C to D. Varying the phase encode amplitude from TR to TR sweeps the trajectory to a different row in *k*-space at each repetition (e.g. dotted trajectory shown).
In the Cartesian raster trajectory, all acquired lines in k-space are equally spaced and parallel.

Spiral filling trajectories may be used in echo-planar imaging (EPI). In a spiral acquisition, central k-space data is acquired at the start of the scan, allowing for very short effective TE times.

In a ‘projection acquisition’, each acquired line follows a radial trajectory from the centre of k-space. The data in a radial line through k-space represents the Fourier transform of a projection of the object in a direction orthogonal to the radial line (‘projection slice theorem’). Acquisition of many radial lines and use of a back-projection algorithm can be used for image reconstruction in manner analogous to CT imaging. Alternatively, radial data can be regridding to conventional Cartesian k-space and a Fourier transform used to recover the image from k-space. Variants on the projection technique were used in the first MR imaging schemes, prior to the now more prevalent spin-warp approach. Radial filling incurs a time penalty of a factor of π relative to Cartesian filling for the same image resolution.

In a PROPELLER sequence, the k-space trajectory is a combination of Cartesian and radial type filling. In a single TR, several lines of data parallel to a radial line through the centre k-space are acquired in a fast spin-echo type acquisition. At the next TR, the direction of acquisition is rotated and a new ‘blade’ of data is acquired. The centre of k-space is acquired for each blade and this data redundancy can be used to correct for rigid motion. PROPELLER is useful in the correction of motion artefact in the head.


**kV meter**

(Diagnostic Radiology) The high voltage supplied to the x-ray tube is measured by a kV meter. There are various kV measuring devices.

The direct kV measuring device uses a special high-voltage divider, which provides low voltage proportional to the high voltage and this way allows the use of a low-voltage meter. The insulation between the high-voltage circuit and the measuring device often uses opto-couplers.

The non-invasive kV device (Figure K.9) usually measures the kV peak (kVP meter). It makes indirect measurement of the kVP by assessing the peak energy in the x-ray spectrum. This device has two photo diodes as detectors of the x-ray radiation. Before reaching the detectors, the x-rays beam is attenuated by special metal filters (different for each detector). This way the two detectors produce different signals, depending on the x-ray beam maximal energy and the attenuation by the filters. Both signals pass through a differential amplifier, which produces a resulting signal. The latter carries information about the maximal x-ray beam energy (kVP). A simpler device would have two copper filters with different thicknesses. More specific kVP meters (used in mammography, CT, etc.) have filters made of materials with specific K absorption edge -- for example molybdenum, rhodium, erbium, platinum, etc.

While the non-invasive kV meter seems like a simple device, the diagnostic physicist should be aware of some of the issues associated with its use. These meters have a specified tolerance, usually 1–2 kV, so the physicist should not make recommendations for adjustment of the equipment, which are more precise than the precision of the meter. kV meters are affected by the angle between the radiation beam and the filter pack, hence the location and tilt of the kV meter are specified by the manufacturer.

As with all other equipment, the meter should undergo an acceptance testing procedure and should be calibrated as recommended by the manufacturer.

**kV selector**

(Diagnostic Radiology) The kV selector is part of the high-voltage control circuit of the high-voltage generator (HVG) of an x-ray equipment. In classical HVG, the kV selector normally uses one or two (broad and fine) switches to select different number of turns from autotransformer -- i.e. different input voltages to the high-voltage transformer (HVT). A gliding graphite roll can also be used instead of switches (see the diagram in the article High-voltage generator).

Contemporary x-ray equipment with high-frequency generator control the kV by varying the frequency of the DC–AC converter. In this case, the kV selector sets the necessary frequency, feeding the high-voltage ferrite transformer (see the article on High-frequency generator).

During the quality control procedure, the kV settings are normally measured indirectly by a kVP meter. A good equipment will have kV with error of the accuracy on the order of 1%. Usually, error above 5% is an indicator of problem in the kV control circuit.

**Related Articles:** High-voltage generator, High-frequency generator, High-voltage control device

**kVP**

(Diagnostic Radiology) See Peak kilovoltage (kVP)

**KZK equation**

(Ultrasound) The KZK equation, or Khokhlov–Zabolotskaya–Kuznetsov equation, is a wave equation that takes into account non-linear propagation, and also diffraction and non-linearity. The equation can be considered a parabolic (one-way) wave equation including a term describing the losses, and quadratic source term for the non-linear contribution. The equation can be solved using techniques that can be divided into three categories: frequency-domain, time-domain, and combined time- and frequency-domain methods. The so-called Bergen code is an example, which is available for download on the Internet. The solutions are computationally intensive, as for each spatial grid point, all intervening beam shapes must be calculated first so that the required amount of cumulative waveform distortion can develop. Other types of wave equations that describe non-linear propagation are one-dimensional diffusion (Burger’s), and full-wave equations (Westervelt).


**FIGURE K.9** Typical non-invasive kVP meter (on the right), with a set of different filters that have to cover the two detectors. The filter pack in the lower left corner is upside down, showing the two metal filters.
Labelling
(Nuclear Medicine) Labelling refers to the process of attaching a radioisotope to a pharmaceutical, also called tracer. The resulting compound is referred to as a radiopharmaceutical. One example is the labelling of 18F to FDG (fluorodeoxyglucose). For radioisotopes routinely used in clinical setting, like 99mTc, there exists a number of labelling kits where each kit is biochemically engineered to target a specific biological process.

Labelling efficiency
(Nuclear Medicine) The fraction of radioisotopes successfully labelled to a targeting compound is described by the labelling efficiency. For many 99mTc pharmaceuticals the labelling efficiency can reach >95% if the preparations are performed in accordance with the manufacturers’ instructions.

Labelling yield
(Nuclear Medicine) The term ‘labelling yield’ is used as a synonym of labelling efficiency. See Labelling efficiency.

Lactate
(Magnetic Resonance) Lactate (lactic acid; Lac) is a chemical compound that features in proton (1H) NMR spectra of the brain. The principal resonance seen is a characteristic doublet at 1.32 ppm due to methyl protons. There is also a quartet at 4.10 ppm due to –CH protons, close to the strong water peak (to observe the CH-quartet in in vivo spectroscopy would be a quality criterion [Figure L.1]).

It is often difficult to identify the 1.32 ppm resonance unequivocally, because contaminating lipid signal appears in the same region of the spectrum. Simple spectral editing can be used to identify the Lac resonance, since, because of scalar coupling, the doublet is 180° out of phase with the rest of the spectrum in spectra collected with an echo time of 136 ms and in phase at 272 ms. Alternatively, a double-quantum filter may be used to eliminate also lipid contamination, for example, in skeletal muscle and to follow the dynamic behaviour of Lac build-up and recovery.

The concentration of lactate in the normal human brain is very low (≈0.5 mM), so the resonance is not visible in healthy subjects. However, the peak increases in size dramatically in hypoxia/ischaemia, reflecting generation of lactate as the end product of anaerobic metabolism. Thus observation of significant levels of Lac in the brain is indicative of abnormal energy metabolism due either to hypoxia (e.g. within tumours or following ischaemic stroke) or to mitochondrial abnormalities.

Related Articles: Editing, Spectral, Magnetic coupling, Magnetic resonance spectroscopy


Lamb wave
(Ultrasound) A lamb wave, or plate wave, is an ultrasound wave, similar to a surface wave propagating in a thin plate with dimensions comparable with the wavelength. Lamb waves propagate in the same direction as the plate surface. Lamb waves are used in non-destructive testing of, for example steel plates and wires.

Related Articles: Longitudinal wave, Transversal wave, Surface wave

Lambert–Beer’s law
(Radiation Protection) The optical radiation passing through matter interacts with its molecules due to absorption and scattering. So the incident radiation fluence $\Phi$ (number of photons per unit area) will decrease $(\Phi - d\Phi)$ after passing the sample of thickness $dx$ (Figure L.2).

If the sample is a homogenous solution of the molar concentration $c$, the decrease of fluence, $d\Phi$, can be expressed as

$$-d\Phi = \sigma \Phi N x c \; dx$$

where $\sigma$ is the cross section (probability) of interaction $N$ is the Avogadro’s number and after integration the expression

$$\Phi = \Phi_0 e^{-\alpha x}$$

This is the so-called Lambert–Beer’s law. It is usually presented in the following form:

$$\Phi = \Phi_0 10^{-\epsilon x}$$

where $\epsilon$ is molar decadic extinction coefficient (mol/dm³).

FIGURE L.1 Molecular structure of lactate.

FIGURE L.2 Attenuation of an optical radiation in the sample of a thickness $dx$. 
The Lambert–Beer’s law is used to determine the ionising radiation dose with Frick’s dosimeter. It is named after the Swiss physicist Johann Heinrich Lambert (1728–1777) and the German physicist August Beer (1825–1863).

**Related Article:** Frick dosimeter


**Laminar flow**

(Ultrasound; General) Laminar flow describes flow that moves in layers. In a tube these layers are parallel to the tube walls. For constant flow in a rigid parallel circular tube, the very thin layer of fluid in contact with the wall does not move and there are progressively higher velocities towards the centre of the tube where the velocity is maximum. There is no mixing of flow between the laminae; if dye is introduced into such flow it can be seen to travel in a distinct streamline not mixing with adjacent flow (Figure L.3).

The flow profile describes the shape of the velocity vectors across the tube. With constant steady flow in a tube with constant circular cross section, the flow profile becomes parabolic. In this particular case, peak velocity is twice the mean velocity in the tube. This is not the case in large arteries which have pulsatile flow and tapering vessels, but can be a useful approximation in some circumstances. Flow is largely laminar in the normal circulation with disturbed or turbulent flow found in some disease states. The flow profiles in arteries with laminar pulsatile flow are described by the Womersley equations (Figure L.4).

**Related Articles:** Reynolds number, Turbulent flow

**Lanthanides**

(General) The elements in the periodic table that have atomic numbers between 57 (lanthanum) and 71 (lutetium) inclusive are known as lanthanides. The name is sometimes used synonymously with the term ‘rare earths’, although the latter includes scandium and yttrium, and excludes the lanthanide promethium.

Lanthanides are silvery white metals which oxidise when exposed to air. Their f shells contain between 1 and 14 electrons. They have the unusual property that their atomic radius reduces with successive filling of the f shell.

**Use in Medicine:** Solid state lasers: Lanthanides are primarily used in solid state lasers, where they provide the impurities necessary for light amplification within the gain medium, which can be a crystal, glass or ceramic. A common laser gain medium for medical lasers is the yttrium aluminium garnet (YAG) crystal, which can be doped with small quantities of various lanthanides. Some lanthanide-doped YAG lasers are listed next, along with their common medical applications.

<table>
<thead>
<tr>
<th>Lanthanide</th>
<th>Use in Medicine</th>
</tr>
</thead>
<tbody>
<tr>
<td>Nd³⁺:YAG</td>
<td>Ophthalmology, dermatology, dentistry</td>
</tr>
<tr>
<td>Yb³⁺:YAG</td>
<td>Laser scalpel</td>
</tr>
<tr>
<td>Er³⁺:YAG</td>
<td>Dentistry, cosmetic surgery, non-invasive blood sugar monitoring</td>
</tr>
<tr>
<td>Ho³⁺:YAG</td>
<td>Dentistry, orthopaedic surgery, kidney stone and tumour ablation</td>
</tr>
</tbody>
</table>

Nuclear medicine imaging radionuclide: Among the lanthanides is gadolinium, whose radioactive isotope ¹⁵³Gd is widely used in nuclear medicine as a transmission source for quality control and attenuation correction.

**Lanthanum oxybromide in intensifying screen**

(Diagnostic Radiology) See Rare earth screen

**Laplace transform**

(Nuclear Medicine) The Laplace transform is an integral transform used to solve physical problems. The Laplace transform is used to transform an unsolvable differential equation into a solvable algebraic expression. See [LaplaceTransform.html](http://mathworld.wolfram.com/LaplaceTransform.html)

**Larmor equation**

(Magnetic Resonance) Due to the torque exerted by a magnetic field on a magnetic dipole, it undergoes a precession when oriented orthogonal to the field. The Larmor equation states the proportionality between the angular frequency of precessing transverse magnetisation and the flux density B

\[
\omega = \gamma |B|
\]

where

- \(\omega\) is called Larmor frequency
- \(\gamma\) the gyromagnetic ratio

It appears as part of the Bloch equations.

**Related Articles:** Gyromagnetic ratio, Bloch equation

**Laser film printer**

(Diagnostic Radiology) The laser film printer (also known as dry laser imager or laser camera) produces a copy of a digital image on a special type of film (often transparent film, similar to x-ray film). The laser film printer is used to produce hard copies of images from various digital imaging systems, as CT scanners, digital radiography, MR, ultrasound, gamma camera, etc.

The laser film printer uses direct laser scanning (projecting the image) onto the film. The device often uses infrared solid state laser with very small laser spot spacing (e.g. 50μm). The initial laser source intensity is constant, but before reaching the film this intensity is changed by a laser modulator to produce intensities, responding to the grey levels of the pixels in the digital image. The modulated...
laser beam scans the film and exposes it to produce an accurate film copy of the digital image. After the exposure the film is developed.

The film for a laser film printer usually contains cubic grains and not tabular grains (as in x-ray screen-film); however, the printer can also work with normal x-ray film and wet-type film processing. Variations in image quality resulting from less than optimal wet film development are potential problems. A newly developed thermographic film developer for laser films (without liquid powdered chemicals) is environmentally better and reduces operating costs. X-ray film with dry processing methods (used with some laser film printers):

1. Adherographic: The film has a laser-sensitive adhesive layer plus imaging layer (carbon particles, pells), both sandwiched between two polyester sheets. When the laser beam scans the dry-film it causes the adhesive layer to take carbon and stick it to the polyester sheet. As a result there are two sheets with positive and negative image. The first is coated and used as film, the other is disposed. The adhesion process is binary and the grey tone (nuance) is produced by dithering. Normally a cell of \(16 \times 16\) pells makes a pixel with 256 grey levels. This requires very thin laser and small pells (5 \(\mu\)m): \(16 \times 5 = 80\) \(\mu\)m pixel, what will produce image resolution of 6.25lp/mm.

2. Thermal: The film emulsion is a combination of silver bhenate and silver halide coated onto polyester. The scanning laser beam triggers ‘thermal developing process' producing a ‘true' greyscale. However there is no fixer – that is, the undeveloped silver halide crystals remain on the film, what makes it thermally unstable.

These laser film printers could produce an image with less grey levels (i.e. less contrast), compared with normal x-ray film. However this is not necessarily a specific disadvantage, because the digital image comes from a digital imaging system, where the ‘window' technique has to be used to obtain optimum contrast by the operator.

The throughput of a printer with wet processing is high (usually more than 100 films per hour). For films with thermal/adherographic dry processing the time for producing one film is much longer (often 30–45 s). The image quality depends on the laser spot and is selectable (e.g. producing 50 or 100\(\mu\)m pixel size with 12 bit greyscale resolution). The printer also has automatic density control and image interpolation technology. The laser film printer (laser imager) allows PACS connection, what makes it very useful for large hospital information systems.

**Laser interferometry**

(*Ultrasound*) A laser interferometry system is used for accurate high resolution (temporal and spatial) displacement measurements. At the National Physical Laboratory NPL, UK, such a system is used as a primary standard instrument for calibrating hydrophones. The system includes a laser, a photodetector, mirror, beamsplitter, thin membrane and a water tank, *Figure L.5*. The equipment is mounted on a table equipped with vibration damping.

A thin acoustically transparent membrane will move in unison with the surrounding medium. When placed in an ultrasound field it will therefore follow the particle displacement. This movement can be calculated from the measurements of the phase difference between the fixed optical reference beam (reflected on the mirror) and the beam, which has been reflected on the moving membrane. When the amplitude of the particle displacement \(s_0\) is known the pressure amplitude \(p\) can be calculated using the relationship \(p = Z \omega s_0\).

**Related Articles:** Displacement, Hydrophone.

![Figure L.5 Laser interferometry. The beam splitter splits the laser beam into two parts. The reference beam is reflected on the mirror and the other part is reflected on the membrane, which is moving in unison with the ultrasound field. The varying phase difference between the two beams is detected by the photodetector. (Graphs courtesy of EMIT project, www.emerald2.eu)](image)

**Laser localisers**

(*Radiotherapy*) In external beam radiotherapy treatment it is important to relate the internal treatment target to the delivery system. A key component of this is to mark the patient’s skin with tattoos that are aligned with the delivery coordinate system using the treatment room lasers. These are known as skin reference marks. Often two lateral and one anterior tattoos and lasers are used.

The first stage of the treatment process involves imaging the patient, for example with CT. The tattoos are often drawn at this point. At treatment simulation and each treatment visit, localisation lasers in each room are used to set the patient up. Often the set-up is further refined using imaging.

**Abbreviation:** CT = Computed tomography

**Related Articles:** Skin reference marks, Treatment verification

**Late response of normal tissue**

(*Radiotherapy*) Radiation treatment inevitably affects normal tissue and so may cause radiation-induced adverse effects. These effects are usually divided into *early* and *late* reactions. Traditionally, reactions that occur more than 90 days after the start of treatment are classified as late effects. For further details on the response of normal tissue to radiation see the articles on *Adverse effects, Dose response model and Tolerance*.

**Related Articles:** Adverse effect, Dose response model, Tolerance

**Latent image**

(*Diagnostic Radiology*) Latent image (hidden image) is a term used in photography to describe the invisible changes in the exposed film before its development. These changes are in fact clusters of metallic silver atoms formed on the silver halide crystals after exposure with light. As it is well known the size of the cluster is proportional to the intensity of the light, thus forming an invisible pattern – latent image. The ‘latent image' term has found stable place in x-ray radiography and other image formation mechanisms. For example, the pattern of absorbed x-rays, formed after these had passed through the object/patient (before exposing the film/detector) is also called ‘latent image'. Sometimes this latent image is also called modulated (by the object) x-ray beam.

**Related Article:** X-rays, X-ray film

**Latent period**

(*Radiation Protection*) The time interval between a person being exposed to a carcinogen, whether it be a chemical pollutant, ionising radiation or any other factor, and the progression of the disease to diagnosis as a leukaemia or solid cancer is known as the latent period. This period can extend to tens of years, and may make it difficult or impossible to make an association between a particular
exposure event and the resulting cancer when carrying out multifactorial analyses or epidemiological studies on populations.

**Related Articles:** Radiobiological models, Stochastic effects

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**Lateral**

(General) Directional anatomical terms describe the relationship of structures relative to other structures or locations in the body. *Lateral:* Towards the side, away from the mid-line (e.g. the little toe is located at the lateral side of the foot).

See Anatomical relationships

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**Lateral electronic equilibrium**

(Radiotherapy) The condition of transient electronic equilibrium implies that the distance between the point where the dose measurement is performed and the limit or edge of the beam is larger than the maximum range of secondary electrons set into motion by photons. In a photon beam the transient electronic equilibrium is assured on the beam axis only for broad beams. Near the edge of a beam there is a lack of lateral electronic equilibrium thereby making accurate dosimetry very difficult. In Figure L.6 build up curves measured in a polymethylmethacrylate (PMMA) phantom at different depths for a 20 MVXR beam are shown. The diameter of the photon beam ranges from 3 to 38 mm. The lack of lateral electronic equilibrium is evident in beams whose width is less than 38 mm. The dose within the field decreases with depth, the penumbra is broadened and the dose deposition outside the field edge increases.

Dose in small radiation beams is therefore inherently difficult to measure. Clinical examples where regions of electronic non-equilibrium are encountered are intensity modulated radiation therapy (IMRT) where small individual beams (beamlets) of less than 1 cm² are commonly employed to create a predefined dose distribution and functional stereotactic radiosurgery (SRS) where similarly sized beamlets may be employed.

**Related Articles:** Electronic equilibrium, Charged particle disequilibrium


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**Lateral position**

(General) There are a series of terms used to describe the position of an individual when undertaking different imaging examination.

*Lateral:* Standing, sitting or lying down with onside in contact with the equipment couch or stand. For example, erect lateral chest x-ray.

See Patient position

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**Lateral resolution**

(Ultrasound) The lateral resolution of an ultrasound system determines its ability to separate objects in the transverse/lateral direction of the image. It is usually defined as the smallest separation of identical point targets at the same depth in the image plane.

Lateral resolution is dependent on beam width. Beam width itself varies throughout the image and is dependent on the depth, aperture (link) and the focal point of the image. In areas of narrow width, separation will be better than zones where the beam is wide (Figure L.7). In practice the intensity of the beam is highest in the centre and falls towards the edge of the beam so that the resulting image of the point target is complex and the lateral resolution will also depend on power, gain and dynamic range chosen.

Figure L.7 shows how lateral resolution is dependent on beam width. In Figure L.7a, three beams sweep across two pairs of targets. At the focal point depth (A), the beam is narrow and the middle beam does not insonate either target. There is a gap seen in the displayed image (Figure L.7b). In the deeper pair (B), the targets are each insonated by two beams and the echoes merge into one another. At the unfocused depth, echoes from the target are always combined with echoes from adjacent tissue with the result that contrast between target and tissue is reduced.

In most scanners, some control of the lateral resolution is possible by selecting an appropriate focal depth for the target under examination. This optimises the transmitted beam width for a particular depth where lateral resolution will be optimised. The effect is shown in Figure L.8. More recently, new beam-forming techniques have dispensed with user-focusing.

Figure L.8 shows that by altering focal depth (arrow cursor to side of image), lateral resolution is optimised to suit the depth of the target under investigation, in this case wires in an ultrasound phantom. In Figure L.8b, the pins at depth (circled) appear round. When the focus is moved more superficially (Figure L.8a), the deep targets appear wider, and contrast and resolution is reduced.

Lateral resolution may be assessed by examining the spread of point targets in a phantom or by examining the separation of groups of point targets. The images in Figure L.9 show the improvements in lateral resolution demonstrated by a group of targets in a phantom.

Figure L.9 shows ultrasound images from a phantom with nine wire targets, five of which lie in a horizontal line with spacing of...
2, 1, 0.5 and 0.25 mm. In the older scanner (Figure L.9a) only the largest (2 mm) gap is evident, with the other four targets merged into one image. In the newer scanner (Figure L.9b), the wires show improved spatial resolution. There are gaps at 1 mm lateral spacing and some separation of the echoes from 0.5 mm. The last two pins appear merged as one.

**Latitude of film**

(Primary Radiology) The latitude of a radiographic film is the exposure range over which it can record contrast as illustrated in Figure L.10.

The latitude is a design characteristic of film and is taken into account when selecting film for specific clinical applications. For example, chest radiography is usually done with a film with relatively wide latitude.

**Related Article:** Characteristic curve

**Lattice**

(General) A crystal is a solid form of matter which has translational periodicity in three-dimensions in its atomic arrangement. Crystal structures are described in terms of a lattice plus a motif. The lattice is an infinite array of points with identical environments and the motif is the element of structure associated with each lattice point. A vector that joins two lattice points is known as a lattice vector (Figure L.11).

The term lattice is also used in magnetic resonance imaging (MRI) to refer to the magnetic and thermal environment of nuclear spins. In longitudinal, or spin-lattice relaxation energy from the spin system is transferred to the lattice.

**Related Articles:** Magnetic resonance, Spin-lattice relaxation

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**FIGURE L.8** Ultrasound images of phantom illustrating focus effect on lateral resolution. (a) Focus superficial to circled pins. (b) Focus at circled pins.

**FIGURE L.9** Ultrasound images of phantom illustrating the effect of the age of the ultrasound machine. (a) Older ultrasound scanner. (b) Newer ultrasound scanner.

**FIGURE L.10** Latitude of film and contrast. (Graphs courtesy of Sprawls Foundation, www.sprawls.org)
Lead coefficient is 59.7 cm−1. In comparison, the coefficient for copper at this energy is 3.8 cm−1. Due to the attenuating properties of lead it is used as a shielding material in hospital departments where radiation is used, be it in walls around x-ray facilities or shielding around vials and syringes in nuclear medicine departments. Lead glass, which contains lead oxide, is used as a shielding material in hospital departments where radiation is used, be it in walls around x-ray facilities or shielding around vials and syringes in nuclear medicine departments.

### Lead

**(General)**

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Pb</th>
</tr>
</thead>
<tbody>
<tr>
<td>Element category</td>
<td>Group IV metal</td>
</tr>
<tr>
<td>Mass number Z</td>
<td>various</td>
</tr>
<tr>
<td>Atomic number A</td>
<td>82</td>
</tr>
<tr>
<td>Electronic configuration</td>
<td>[Xe] 4f14 5d10 6s2 6p2</td>
</tr>
<tr>
<td>Melting point</td>
<td>600.6 K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>2022 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>11.34 g/cm³</td>
</tr>
</tbody>
</table>

**Lead** (chemical symbol: Pb from the Latin ‘plumbum’) is a metal in Group IV of the periodic table. At room temperature and pressure, lead is a dense solid and possesses a cubic close-packed structure (cpp). It is a soft, malleable material, which lends it to use in construction. Lead is a poor conductor of heat and electricity compared with other metals. Lead is highly resistant to corrosion and is therefore often used to contain corrosive materials such as strong acids. A flame test identifies lead with a whitish-blue colour.

**Isotopes:** The four stable isotopes of lead are 204Pb, 206Pb, 207Pb and 208Pb. In addition, there are numerous unstable isotopes.

**Medical Applications:** The linear attenuation coefficients for lead are high. For example, for 100 keV γ-rays the linear attenuation coefficient is 59.7 cm⁻¹. In comparison, the coefficient for copper at this energy is 3.8 cm⁻¹. Due to the attenuating properties of lead it is used as a shielding material in hospital departments where radiation is used, be it in walls around x-ray facilities or shielding around vials and syringes in nuclear medicine departments. Lead glass, which contains lead oxide, is used as a shielding material in situations where the material needs to be transparent, for example x-ray control rooms or eyewear. In addition, lead aprons are worn to protect staff working in x-ray rooms, particularly where fluoroscopic procedures are performed. Gonad shields are also made from lead. Furthermore, lead forms part of the material lead zirconate titanate (PZT), a material which displays a marked piezoelectric effect. As such, PZT it is used in the construction of ultrasound transducers.

**Related Articles:** Linear attenuation coefficient, Lead glass

**Lead apron**

**(Radiation Protection)** Registrants and licensees shall ensure that workers are provided with suitable and adequate personal protective equipment which meets any relevant regulations or standards. Protective equipment includes lead aprons, thyroid protectors, protective eyewear and gloves. In practice, the need for these protective devices should be established by the qualified expert on radiation protection or the radiation protection officer.

Employers, registrants and licensees shall ensure that all personal protective equipment be maintained in proper conditions and be tested at regular intervals.

Gowns, aprons and thyroid protectors are usually made of material, such as vinyl, which contains lead. Aprons should be equivalent to at least 0.25 mm Pb, if the x-ray equipment operates up to 100kV and at least 0.35 mm Pb if it operates above 100kV. The aprons should be kept properly, hanging and not folded, and checked regularly, under fluoroscopy.

Several models of lead aprons are available in order to better fit the size of the worker and the activity. In situations where occupational exposure is expected to be high, such as during interventional radiology procedures, the lead apron should be of the wrap-around type. A combination of vest and skirt is also possible. In cases when the operator is always facing the radiation source, the apron may be open at the back or with less lead at the back, in order to reduce the weight and cost.


**Lead content**

**(Radiation Protection)** Materials used for shielding are often quoted in lead equivalent (or content) which is the thickness of lead which will provide the same amount of shielding. Lead is one of the most used materials for radiation shielding of x-rays and γ-rays. The advantage of lead is that it is easily shaped into sheets and interlocking bricks. Therefore it is easy to insert lead into wooden panels for the walls (adding protection) or mobile shielding barriers. Lead is also used for staff personal protective devices (lead curtains, aprons, gloves, etc.), patient protection devices (gonad shield, etc.) or for the shaping of radiation beams (diagnostic and therapy applications).

The density of lead is approximately 11 times greater than water. As an indication 4 cm of lead attenuate a Co-60 beam to 1/10th of the unshielded value (without considering geometric and beam effects). When other materials are used for shielding, their respective protective values are usually given in terms of lead equivalent. The shielding effectiveness (lead content) of a radiation barrier is given in terms of lead equivalent thickness in mm. There are also some negative aspects related to the use of lead, as for example that the sheets might slide, due to the high weight (there is need of regular checking) and that the cost is rather high. Therefore, when possible, as in the case of building walls, etc. concrete is the most recommended shielding material (there are also special kinds of concrete).


**Lead drapes**

**(Radiation Protection)** Lead drapes or curtains, usually in the shape of stripes of lead covered with plastic material are used to protect the worker who is standing near the patient and the x-ray
tube. For example, a typical setting for fluoroscopy investigation (with the x-ray tube under the patient and the workers nearby) requires this type of protection to be put between the tube and the workers. The stripes allow movements and adjustments. It is important that the stripes are a little overlapping each other, in order to ensure good protection. Regular control should be made as lead might slide inside and reduce the protection.

**Lead equivalent**
*(Radiation Protection)* See Lead content

**Lead glass**
*(Radiation Protection)* Lead glass is a form of glass that has an amount of lead in the form of lead oxide. Depending on the amount of lead oxide the atomic number of the glass increases, but at the same time maintains the transparency of the glass. Lead glass usually has an attenuation factor of 1.8–3.2 mm lead equivalent.

Lead glass is used to provide a view port (window) on a shield for ionising radiation, for example a window on a movable lead shield, eyewear, etc.

**Lead glasses (eyewear)**
*(Radiation Protection)* The human eye is particularly sensitive to damage caused by ionising radiation (potentially causing cataracts and other opacities).

There are some work activities with ionising radiation that involve workers standing in close proximity to a radiation source, for example during interventional radiology/fluoroscopy investigations. In such cases, where it is determined (by a risk assessment) that there is a risk of a significant radiation dose to the eyes, the employer is required to provide protection in the form of leaded eyewear.

Ideally the eyewear should be shaped such that the sides are protected as much as the front face. However the main problem with protective eyewear is that to provide reasonable protection it may be bulky or heavy such that wearing the eyewear for long periods feels uncomfortable.

**Lead gloves**
*(Radiation Protection)* Registrants and licensees shall ensure that workers are provided with suitable and adequate personal protective equipment which meets any relevant regulations or standards. Protective equipment includes lead aprons, thyroid protectors, protective eyewear and gloves. In practice, the need for these protective devices should be established by the qualified expert or the radiation protection officer.

Employers, registrants and licensees shall ensure that all personal protective equipment be maintained in proper conditions and be tested at regular intervals.

Lead gloves are used to protect the hands, but the weight might limit the movements. In particular gauntlets are very heavy gloves and should be used only when appropriate (e.g. holding patients). Special light gloves are available for fluoroscopy.


**Lead protection**
*(Radiation Protection)* Lead is commonly used to shield against ionising radiation, particularly in the energy ranges where the photoelectric effect is predominant (e.g. x-ray energies below 100kV). Attenuation of photons through the photoelectric effect is proportional to \( Z^2 \); hence high \( Z \) materials such as lead cause high attenuation.

Lead is used as wall shielding for x-ray rooms. Usually a sheet of lead between 1.3 mm (Code 3) and 2.65 mm (Code 6) sandwiched between plywood (to provide rigidity) is used.

Lead is also incorporated into other materials such as glass or rubber to provide shielding for other purposes. Lead glass panels are used extensively in x-ray rooms to allow the operator to view the patient during procedures. Radiologists or cardiologists making extensive use of fluoroscopy may wear lead impregnated glasses, providing up to 0.5 mm lead protection for the eyes.

Lead rubber is extensively used in the manufacture of x-ray protective clothing, for example aprons, gloves, thyroid collars. These will normally provide between 0.25 and 0.5 mm lead protection. This is sufficient to protect against scattered (low energy) x-radiation.

Lead is also incorporated into rubber or plastic gonad shields. These articles normally have up to 0.5 mm lead thickness.

**Related Article:** Shielding

**Leak test**
*(Radiation Protection)* A leak test is an assessment of the integrity of encapsulation of a sealed radioactive source, or from the housing of an x-ray tube. Leak testing of sealed sources and x-ray tubes should each be considered apart, because they involve separate approaches.

Leak testing of sealed radioactive sources is carried out on an annual basis by a wipe test of the source in question. The aim is to determine whether there is any leakage of the radioactive substance from the casing or encapsulation, which would be detected as contamination on the swab. There are statutory limits on the maximum activity detected using the wipe test method.

Leak testing of an x-ray tube is performed by measuring dose rates around a completely closed (i.e. collimated) tube-housing, using sensitive ionisation chambers. Readings are taken at as many orthogonal directions around the housing as possible (at least one reading per direction, requiring a minimum of six). X-ray cassettes may also be employed to further investigate the location of leakage from the tube housing. The maximum possible dose rates to bystanders can be calculated from the measured leakage, and compared to statutory limits. This is particularly relevant with mobile and dental x-ray equipment where operators and/or public may be nearby. X-ray tube leak tests are performed as part of commissioning tests (including installation of a replacement x-ray tube).

**Related Articles:** Wipe test, Radiotherapy, Brachytherapy

**Leak test**
*(Radiotherapy)* Sealed sources used in brachytherapy must be tested for leakage and contamination. In general, yearly checks are required. Note that national regulations may vary in their requirements.

For HDR and PDR sources (remote afterloading devices), which are replaced four times a year, the leakage test performed by the manufacturer for each source is stated in the accompanying source certificates (requirement: leakage and contamination test <0.185 kBq). The recommended contamination test for the hospital is an applicator test, performed by placing an applicator in a well-type counter to detect any photon emitting contamination. Contamination of the dummy source can be tested by a wipe test, as it is normally possible to drive the dummy source out using manual control. Note that the dummy source is also replaced at regular intervals.
For LDR and MDR sources (afterloading and manual loading), the recommendation is to check contamination of sources or leakage of radioactive material using a wipe test. Instruments used to cut 192Ir-wires should also be checked.

For permanent seed implants, the source certificate states (Oncura 125-I RAPID Strands): ‘All seeds have passed a leakage and contamination test showing less than 0.185kBq, 0.005μCi of removable Iodine-125 activity’.

Further reading can be found in the ESTRO Booklet No. 8

Abbreviations: ESTRO = European Society for Therapeutic Radiology and Oncology, HDR = High dose rate and PDR = Pulsed dose rate.

Related Article: Dummy source


Leakage current

(General) Leakage current is an unwanted flow (leak) of current along a path different from that intended. In electric circuits and components, it is mainly due to insufficient or faulty insulation (e.g. leakage current of a capacitor). In electronic circuits, it is an inherent characteristic of semiconductor devices (e.g. leakage current in analogue switches).

Leakage radiation

(Radiation Protection) Leakage radiation results from the lack of integrity of radiation shielding. This may occur from cracks in the housing of an x-ray tube. Leakage radiation may also refer to radiation penetrating through service-ducts in walls, gaps between shielded doors in x-ray facilities, etc. This results in higher than expected dose rates adjacent to the gap in the shielding, with potential consequences for harm of persons in that area.

Leakage radiation from facilities is checked at construction as part of general tests to assess actual wall-shielding against a build specification. Leakage radiation from x-ray units is assessed at unit commissioning and installation of new x-ray tubes, by means of a leak test.

Related Article: Leak test

Leakage radiation

(Radiotherapy) Leakage radiation is the ionising radiation which has passed through the protective shielding of a radiation source. It is important to reduce this leakage to protect the patient from unwanted radiation and to help reduce the necessary amount of additional treatment room shielding. It is minimised by the use of significant amounts of protective shielding (lead, tungsten, etc.) in the head of the linac.

The leakage radiation from a linac is measured in air with the collimator jaws closed to block the primary beam. A chamber with an appropriate build-up cap is placed at a distance of 1 m from the target and also at 1 m from the target through the linac. Measurements are also made in the plane of the patient at isocentre height (for details see IEC 60601-2-1). It is possible to compare this to a measurement made of the primary beam sized 10 × 10 cm2 at the isocentre for 100 MU. A sufficient reading for the leakage measurements can be achieved by setting 1000 MU and then dividing the answer by 10 for comparison with the open field reading. The average leakage measurements for the region shielded by the collimators should be less than 0.5% of the open field and the average leakage for the region surrounding the accelerator and waveguide should be less than 0.1% of the open field.

A simple test to confirm the coverage provided by the lead within the head of the linac, in addition to local ‘hot spots’, can be performed by wrapping the head of the linac with radiographic film. The jaws are fully closed and 1000MU delivered. To allow comparison some known control films should be generated by irradiating film with 10 and 100MUs placed at the isocentre with 1 cm build-up and a field size of 10 × 10 cm2. A comparison of the optical density readings can then be performed to assess the integrity of the shielding.

For high energy linacs (energy greater than 10MV) it is also necessary to assess the level of neutron leakage when designing treatment rooms.

Related Articles: Boron neutron capture, Collimation, Treatment head, Maze, Secondary barrier, Tongue and groove leakage


Lens

(Ultrasound) Mechanical focusing of an ultrasound beam is possible with an acoustic lens or with a curved transducer element. Figure L.12. An acoustic lens works in a similar way to an optical lens. However, in the optical case speed of light in the lens material is always lower than speed of light in air. In the acoustical case, speed of sound in the lens material can be either higher or lower than that for tissue (1540m/s). Convex (speed of sound lower than tissue) or concave l (speed of sound higher than tissue) lenses can be used for focusing, Figure L.12.

The focal distance, F, is determined by the radius of curvature (=F) in the case of a curved transducer element case and by both the curvature and speed of sound in the case of focusing with a lens. Focusing can only be achieved within the unfocused transducer’s near field. The beamwidth WF at a distance F depends on the transducer diameter D and the frequency. It can be approximated to WF = FKdD.

The focal zone is defined as the region where the beam width W < 2WF.

Mechanical focusing is used in single element transducers, mechanical sector scanners and in the elevation plane for conven-

**LET**
*(Radiation Protection) See Linear energy transfer (LET)*

**Lethal dose**
*(Radiation Protection)* In general, the term lethal dose refers to the radiation dose that will cause the death of the individual that receives it. The 50% lethal dose (LD₉₀), defined as the dose that causes a mortality rate of 50% in an experimental group within a specified period of time, has been adopted as an end-point for scoring radiation death.

The effect of a dose of radiation will depend on its magnitude, the proportion and/or the part of the body exposed. Single doses up to 10 Gy and multiple doses of 2–3 Gy are routinely delivered to a limited area of the body in palliative and radical radiotherapy treatment. However, a single dose of only 2.5–5 Gy to the whole body may be sufficient to cause death. At such doses, death is caused by radiation damage to the haematopoietic system.

There is a time delay between the radiation damage and the onset of symptoms since it is the mitotically active precursor cells (precursor cells are stem cells that have developed to the stage where they are committed to forming a particular kind of new blood cell) that are sterilised by the radiation. Hence it is the subsequent supply of mature red blood cells, white blood cells and platelets that is reduced, and so it is only when the current circulating mature cells begin to die off that the inability of the depleted precursor cell population to replace them becomes apparent.

In humans, the peak incidence of death from haematological death occurs at about 30 days after exposure but deaths continue for up to 60 days. Therefore LD₉₀ estimates for humans are usually expressed as the LD₉₀₆₀. Studies of patients who have received total body irradiation for radiotherapy and those involved in radiation accidents/incidents such as the victims of Hiroshima and Chernobyl have attempted to estimate the value of LD₉₀₆₀ and this is generally quoted at 4 Gy. However, there are many factors that influence the response of an individual within a population. For example, females appear generally to have a greater dose tolerance than males as do young adults compared with the very young and elderly.

**Abbreviations:** LD₉₀ = 50% lethal dose and LD₉₀₆₀ = 50% lethal dose measured at 60 days from time of exposure.

**Related Articles:** Cell cycle, Fractionation, Palliative treatment, Radiosensitivity, Tolerance, Total body irradiation


**Light field**
*(Radiotherapy)* It is essential to be able to confirm the settings to be used for creating the treatment field shape and also to visualise the treatment field on the patient’s skin. To achieve this, a light field which exactly mimics the radiation field to be delivered can be used. The linear accelerator contains a light bulb within the head and a mirror to shine the field through the jaws. The bulb is aligned such that the light field will mimic the radiation field to be delivered to within 1 mm on each jaw. It is important that the distance from the bulb to the mirror is the same as the distance from the source to the mirror. The light field also allows a projection of the crosshairs to be displayed and used to confirm the crosshair stability with collimator rotation. It should be possible to achieve crosshair rotational walkout to be less than 0.5 mm.

It is important that the coincidence of the light and radiation fields is checked on a regular basis over a range of different field sizes. The most common method to do this is using a radiographic film; however with the increase in digital techniques other methods are becoming popular such as using electronic portal imaging devices. A piece of radiographic film is marked by pen according to the field indicated by the light field, then without moving the film or changing the field size the film is irradiated. It is then possible to compare the field size according to both the radiation (readout using a densitometer) and the light field (measured with a ruler), and the coincidence between the two.

The calibration of the secondary collimators (jaws) can also be checked by using the light field projection onto a piece of graph paper positioned at the isocentre plane.

**Related Articles:** Collimator, Crosshairs, Optical distance indicator

**Light guide**
*(Nuclear Medicine)* Light guide refers to the connection between the crystal and the photomultiplier tube (PM tube) in a gamma camera. The light photons are generated in the crystal as a result of electrons released in a photon–electron interaction, that is photoelectric effect or Compton interaction. The light guide allows photons to pass from the crystal to the PM tubes where the signal is strengthened. In a gamma camera, several PM tubes are packed close together and connected to a single crystal by one light guide. In early emission imaging the light guide was kept relatively wide in order to attain a uniform irradiation of the PM tubes. A wide light guide will have a degenerative effect on the spatial resolution (as illustrated in Figure L.13) because of an increase in deposition point to PM tube distance. However a modern uniformity

**FIGURE L.13** A wide light guide will have a negative impact on the spatial resolutions because of the increase in deposition point to PM tube distance.
Light localiser

for sodium iodine with thallium impurities 20 times the bandgap to create an electron–hole pair. As an example, material. It takes for a number of materials an average of three photons that are emitted per unit radiation energy deposited in the sodium iodine detector, 5 × 10^4 electron hole pairs would be created. The light yield is thus close to 1 light photon per electron–hole pair.

Related Article: Photomultiplier (PM) tube

Light localiser

(Diagnostic Radiology) A device used to adjust the radiation field/beam at an exact place. Radiographic devices use a diaphragm (light beam diaphragm) to mimic the x-ray field with light and adjust the exposure field over specific part of the patient. Scanning devices (as CT scanners) use laser beam mimicking the x-ray scanning beam to set up the exact scanning plane. Precise positioning of the light localiser is checked during quality control procedures (Figure L.14).

Related Article: Diaphragm collimator

Light yield in scintillation detectors

(Nuclear Medicine) Light yield is an important parameter of a scintillating material. It is the measure of the number of light photons that are emitted per unit radiation energy deposited in the material. It takes for a number of materials an average of three times the bandgap to create an electron–hole pair. As an example, for sodium iodine with thallium impurities 20 eV is required to create an electron hole pair. If 1 MeV were deposited in the sodium iodine detector, 5 × 10^4 electron hole pairs would be created. The total number of light photons created from 1 MeV is 4 × 10^4 with 3 eV each. The yield is thus close to 1 light photon per electron–hole pair.

Related Articles: Inorganic scintillators, Scintillators, NaI(Tl) detector crystal, Bismuth germanate (BGO)


Limitation

(Radiation Protection) The third principle of protection against ionising radiation for workers and members of the public specified by the International Commission on Radiological Protection is limitation.

Once the use of ionising radiation has been justified and optimised the exposure must be limited. The current (ICRP Publication 103) recommendations for exposure limitation are divided into stochastic limits and deterministic limits.

The stochastic limit is 100 mSv effective dose over five years with no more than 50 mSv effective dose in any one year.

Limited angle tomography

(Diagnostic Radiology) Radiographic tomographic procedure (linear classical tomography) in which the angle of motion of the x-ray tube is set to a low value (like 10 angular degrees) to produce relatively thick image slices (as Zonography).

LINAC

(Radiotherapy) See Linear accelerator

Line artefact

(General) Line artefacts cross the image as stripes or solid or dashed lines. In nuclear medicine they have been caused by problems with the analogue to digital converters. In MRI they may be caused by RF leakage or errors in the RF transmission. They can also be caused by problems with the image memory array.

Related Article: Artefact
Line focus principle
(Diagnostic Radiology) The line focus principle, first described by Dr. O. Goetze, in 1918, is applied to most x-ray tubes. The usual cathode consists of a helical heated filament mounted in a focusing electrode. The resulting electron beam focused on the anode surface forms a focal spot that is an image of the elongated (line shaped) heated filament. The length is generally the largest dimension of the focal spot and is highly dependent on the angle of the anode surface and the direction from which the focal spot is being observed. For a line focus principle-related diagram see the article Stationary anode.

Related Article: Stationary anode
Further Reading: Goetze, O., DP 370 022. 1918

Line of response (LOR)
(Nuclear Medicine) A registered event is assumed to originate from a decay or annihilation somewhere along a line of response (LOR). The way a LOR is determined differs between PET and a scintillation camera due to the different spatial localisation processes involved. For example, in PET imaging, when two opposite detectors simultaneously register an annihilation photon a LOR is ‘drawn’ between them, that is the event is assumed to have occurred somewhere along the LOR. In a scintillation camera, if the camera uses a parallel-hole collimator, each hole is associated with a LOR that is perpendicular to the detector surface.

Related Articles: Spatial resolution, Full width at half maximum (FWHM)

Line pair
(Diagnostic Radiology) A line pair is one line and one adjacent blank space in a resolution or bar-phantom test object as illustrated in Figure L.15.

For x-ray imaging applications, the lines consist of an absorbing material such as lead. The typical test pattern consists of a series of different-sized lines and spaces, or line pairs.

The size of the lines and spaces is specified as the number of line pairs in a unit length, line pairs per mm (lp/mm, also known as spatial frequency, in analogue to cycles per second – c/s frequency).

Related Articles: Spatial resolution, Detail resolution, Bar phantom

Line scanning
(Ultrasound) The term line scanning refers to images constructed from an ultrasound beam or ultrasound beams. The echoes arise from reflection and scattering along the line of the beam. Single line amplitude scans are described as A-line scans. B-mode scans are constructed from echoes along several lines.

Related Articles: A-lines, B-lines, B-mode

Line source model
(Nuclear Medicine) This refers to the process in nuclear medicine whereby a line source is used to examine the spatial resolution of the imaging system.

A planar image of one or more line sources is acquired under certain specified conditions (e.g. according to NEMA protocol). A profile is then drawn across the source. The full width at half maximum (FWHM) of this curve gives the spatial resolution of the system.

Related Articles: Spatial resolution, Full width at half maximum (FWHM)

Line spread function (LSF)
(Nuclear Medicine) A point spread function (PSF) is a function that describes an image system degrading effect due to inherent limitations. In its basic concept, a point spread function is a 2D (or 3D) image of an infinitesimal small object. In nuclear medicine applications this will be a point source that is much smaller in radius compared to the expected spatial resolution of the system.

Sometimes it can be difficult to work with and prepare a point source. A measurement of the line spread function (LSF) can therefore be the choice. The LSF is the line spread function integrated in either x- or y-direction. In practice a LSF measurement of a line source is made with a scintillation camera. The line source is usually a capillary plastic tube with a very small diameter filled with a radioactive solution. The source should be longer than the tails of the expected PSF. The LSF function is then approximated by a profile through the centre of the imaged line source.

Abbreviations: LSF = Line spread functions and PSF = Point spread functions.

Related Article: Point spread function, MTF

Line voltage
(General) The voltage provided by a power line and measured at the point of use.

Households are usually connected to a single-phase electric power system. Line voltage is measured between the active (phase) and neutral conductor. The nominal value of the line voltage and the nominal frequency of the alternating current differ geographically. In Europe and Asia it is in most countries 230V/50Hz, and in the Americas 115V/60Hz.

In a three-phase electric power system, line voltage is measured between any two active (phase) conductors (and not referred to the neutral conductor). Line voltage in a three-phase system is 1.73 times higher than phase voltage.

Three-phase power systems are used for supplying large power scale customers. Some medical imaging equipment, for example x-ray machines, CT scanners and MRI systems, are usually powered by a three-phase electric power system.

Related Articles: Y-voltage, Star voltage

Linear accelerator
(Radiotherapy) A linear accelerator (LINAC) is the most common megavoltage unit used in radiotherapy to produce MV photon and MeV electron beams at several different energies, see Figure L.16. It is gradually replacing the 60Co and kilovoltage treatment machines due to its increasing range and scope of clinical treatment techniques. It is housed in a gantry that allows the radiation beam to be rotated isocentrically through 360° about the treatment couch.

An electron gun is used to inject electrons into a waveguide, in synchrony with radiofrequency energy from a magnetron or klystron. This accelerates the electrons to high energies and they are
then magnetically steered within the treatment head to either form an electron beam or directed at a target to produce bremsstrahlung radiation, often incorrectly called an x-ray beam.

The structure and main components of a LINAC can be seen in Figure L.17, and are as follows:

- Electron gun
- Accelerating waveguide
- RF power source
- Pulsed modulator
- Water cooling system
- Vacuum system
- Steering and focusing magnets
- Bending magnets
- Treatment head

**Electron Gun and Waveguide:** Electrons are produced in the electron gun by thermionic emission from a heated tungsten cathode and focused into a central stream by accelerating anodes. They pass into a waveguide in which they are accelerated up to speeds close to the speed of light. As the electrons travel down the waveguide, their magnetic fields interact with the RF field (electric field component aligned along the long axis) and they experience a force, accelerating through the tube. By using travelling RF waves, or standing RF waves, the force continues to accelerate the electrons even as they move, producing a high-energy electron beam. An advantage to the standing waveguide is that it produces a higher accelerating gradient per metre, allowing for more compact vertically designed machines that need no bending magnets. Steering and focusing magnetic coils are used along the waveguide to counter the dispersal of the electrons from the central axis. This is all contained within a vacuum system to ensure that the accelerated electrons are not deflected from their path by collisions with gas molecules.

**RF Power Source:** There are two main RF power sources used in LINACs – magnetrons and klystrons. A magnetron is an RF oscillator that extracts energy from electrons in a resonant structure within a magnetic field. A klystron is a tuned RF amplifier, which amplifies an external source of RF to higher power. Klystrons are generally considered to be more reliable than magnetrons, which can be affected by the earth's magnetic field. However they are much larger and need more sophisticated waveguides to transfer the power, as they cannot be mounted within the gantry, like magnetrons.

**Treatment Head:** The target used to create a MV photon beam is made from Tungsten, and is placed in the path of the electron beam. The Bremsstrahlung photons produced by the interactions in the target will be highly peaked along the central axis. To create a clinically useful beam, a flattening conical filter is used to reduce the intensity in the centre. This filter is energy specific. A dual ionisation chamber system is used to monitor energy, flatness and dose.

If an electron beam is required, the photon target is moved out of the path of the electron beam, and is replaced by two scattering foils. These modify the shape and spectrum of the intense pencil beam of electrons to produce a clinically useful beam. Electron beams require additional collimation called applicators to reduce...
the dose from the high scatter that occurs in air. The beam current in electron mode must be reduced by a factor of 100 (along with the frequency of the RF source) from that of x-ray mode to prevent dangerously high dose rates for electrons.

**Abbreviations:** LINAC = Linear accelerator and RF = Radio-frequency.

**Related Articles:** Treatment head, Magnetron, Klystron, Bending magnet, waveguide


### Linear array

*(Ultrasound)* Linear arrays are array transducers in which the elements, typically 64, 128 or 256 are arranged in a straight line (Figure L.18). This image format is rectilinear with the width dictated by the length of the array. Linear arrays are most commonly used for imaging of superficial tissue including breast, thyroid and testes, musculoskeletal applications and peripheral arteries and veins. They typically use frequencies in the range 3–16 MHz depending on the application. For colour flow and pulsed wave spectral Doppler imaging, linear arrays enable electronic beam steering (Figure L.19) which permits an adequate beam/flow angle for peripheral vessels, many of which run parallel or nearly parallel to the skin surface. In some systems, a trapezoid image can be obtained using the outer elements for beam steering in B-mode.

Multi-D or matrix linear arrays are offered by some manufacturers with dynamic focusing in the elevation plane. In these transducers, elements are subdivided across the width of the probe.

**Related Articles:** Matrix array, Curvilinear array, Transducer array

### Linear array transducer

*(Ultrasound)* See Linear array

### Linear attenuation coefficient

*(Radiation Protection)* Attenuation of photonic radiation is an exponential process that can be represented by the equation

\[ I = I_0 e^{-\mu x} \]

where

- \( I_0 \) is the initial beam intensity
- \( I \) is the final beam intensity
- \( x \) is the distance travelled
- \( \mu \) is the attenuation coefficient of the material (absorber)

### Linear (classical) tomography

*(Diagnostic Radiology)* Linear tomography is a radiographic procedure for producing tomograms or images of selected slices within a patient’s body. The process is illustrated in Figure L.20. During the exposure the x-ray tube and the receptor are moved as shown. This motion blurs the anatomical structures both above and below the slice that is being imaged.

**Related Articles:** Attenuation, Mass attenuation coefficient, Atomic attenuation coefficient

The tube and receptor are on opposite end of an arm that pivots about a point that is in the same plane as the slice being imaged. The location of the pivot point relative to the body can be adjusted to produce images in different planes, or at different depths, through the body.

The objective is to blur and reduce the contrast of the anatomical structures above and below the tomographic slice, or plane of cut, without blurring the image in the slice itself. How this is achieved is illustrated in Figure L.21.

The image of an object located in the plane of cut, or tomographic slice, does not move relative to the receptor and is therefore not blurred. However, the image of an object that is not in the plane of cut will move relative to the receptor and will be blurred. An illustration is seen on Figure L.24.

The amount of blurring is determined by both the distance of an object from the plane of cut and the angle through which the tube and receptor moves as illustrated in Figure L.22.

Linear tomography does not produce a slice with a precise thickness in which there is no blurring. As illustrated here, the blurring increases with distance from the plane of cut. The slice thickness in which there is relatively little blurring is determined by the angle. The angle through which the tube and receptor moves is an adjustable factor and is used to set the slice thickness.

The need to produce very thin slices requires high angle of movement (long pathway of the tube and film, Figure L.23). When this is not achievable technologically the x-ray tube and the film move around the object not in a linear pathway, but form a complex curve. This can be a circle (the tube and the film rotate in opposite directions), or other complex curves as hypocycloids, spiral, etc. Due to the fact that in such complex movements the length of the pathway of the tube and the film is much longer than in linear movement, the tomographic slices are very thin. However in this case the patient dose is much higher. With the introduction of computed tomography and other scanning imaging method the importance of the classical linear tomography has rapidly decreased (Figures L.23 and L.24).
Linear dose response curve

(Radiation Protection) The linear dose response curve is just one example of a dose response curve, together with non-threshold dose response curve, non-linear response curve, etc., which may be used either separately or in combination as models to describe the response of the human body to exposure to various types of ionising radiation from both internal and external exposure, and at high and low doses and dose rates.

The current internationally accepted framework for radiation protection is based on a model of potential harm from exposure to ionising radiation called the linear no-threshold model, which assumes a non-threshold linear dose response curve. This model suggests that at any level of received radiation dose, harm may be caused – that is, a cancer may be induced. The risk of harm (stochastic effects) is proportionate to the dose received (i.e. linear with dose with no threshold for the effect). This is described in Figure L.25.

However, although it is hypothesised that the response is linear with dose, and that there is no threshold, there is no strong evidence to suggest that this is indeed the case.

Related Articles: Dose response curve, LNT model

Linear energy transfer (LET)

(Radiation Protection) Linear energy transfer describes the way in which the energy of incident ionising radiation is transferred to the medium through interactions (ionisations) with electrons or atomic nuclei. Ionising radiation with low density or sparse interactions along the tracks of incident photons/particles is called low LET radiation. Conversely, radiation with a high density of interactions along the track of incident photons/particles is called high LET radiation.

Examples of low LET radiation include gamma rays, x-rays and beta particles. Examples of high LET radiation include alpha particles and neutrons. This property of ionising radiation is partly described by the radiation weighting Factor used to calculate equivalent dose.

Related Articles: Energy deposition, Radiation weighting factor, Equivalent dose

Linear gradient

(Magnetic Resonance) See Gradient linearity

Linear no-threshold dose response

(Radiation Protection) The current internationally accepted framework for radiation protection is based on a model of potential harm from exposure to ionising radiation called the linear no-threshold model. This model suggests that at any level of received radiation dose, harm may be caused – that is, a cancer may be induced. The risk of harm (stochastic effects) is proportionate to the dose received (i.e. linear with dose with no threshold for the effect). This is described in Figure L.25.

However, although it is hypothesised that the response is linear with dose, and that there is no threshold, there is no strong evidence to suggest that this is indeed the case.

Related Articles: Linear dose response, Hormesis, Dose response model


Linear no-threshold model

(Radiation Protection) The basis of radiation protection, as described by the International Commission for Radiological Protection (ICRP) is that the stochastic effects of ionising radiation can occur at any level of exposure. In other words, radiation exposure can cause harm to humans even for the smallest dose received.

Furthermore, it is assumed that the level of risk – that is the likelihood of suffering a stochastic effect – is proportional to the dose received.

This hypothesis is known as the linear no-threshold model, and can be described by the non-threshold dose response curve in Figure L.26.

The graph demonstrates a linear (i.e. proportional) relationship between dose and risk of effect, and the intercept with the axes is at the origin – there is no threshold dose below which a human is safe from the risk of harm. However, the epidemiological evidence used to support the LNT model is mainly at high doses/high dose rates, and it is therefore merely an assumption that there is a linear relationship down to the smaller doses experienced in occupational and diagnostic medical exposures.

More recent epidemiological evidence would suggest that the linear no-threshold model may be too simplistic, and that there may even be a beneficial effect of exposure to ionising radiation at low doses – this is termed hormesis – the hormetic effect of radiation exposure.

Related Articles: Hormesis, International Commission for Radiological Protection (ICRP), Stochastic effects


Linear quadratic (LQ) model

(Radiotherapy) The linear quadratic (LQ) model is a dose–response model and is probably the most widely used and the best currently available for describing the form of the cell survival curve. It is a second-order polynomial function with only two adjustable parameters, $\alpha$ and $\beta$, and a zero constant term so that the surviving fraction is equal to 1 at zero dose.

![FIGURE L.26 The linear no-threshold model.](Image 359x69 to 525x173)
The LQ model assumes that there are two components of cell killing by radiation: one proportional to dose and one proportional to the square of the dose. It was first applied as an empirical description of how the survival of cells varied with radiation dose but radiobiological mechanisms were subsequently attached where the linear component is attributed to single-track events and the quadratic component attributed to two-track events (for more details see the article on Alpha beta ratio). The LQ model expression for the cell survival curve is given by Equation L.1 in which the surviving fraction, SF, is the fraction of cells surviving a single dose $D$ and $\alpha$ and $\beta$ are constants.

$$\text{Surviving fraction, } SF = e^{-(\alpha D + \beta D^2)} \quad (L.1)$$

The LQ model is a second-order polynomial function relating the surviving fraction to dose using only two adjustable parameters, $\alpha$ and $\beta$.

The parameters $\alpha$ and $\beta$ (with units of $1/\text{Gy}$ and $1/\text{Gy}^2$ respectively) determine the ‘bendiness’ of the survival curve, see Figure L.27. The ratio $\alpha/\beta$ has the unit Gy, and in a semi-log plot of $SF(D)$ it is the dose where both the linear and the quadratic components of the survival curve are equal. As shown in Figure L.27, the response of cells to densely ionising high-LET radiation (e.g. neutrons and $\alpha$-particles) usually results in a steep, almost exponential survival curve (see also the article on Relative biological effectiveness). In the LQ model description this would be explained by a high $\alpha/\beta$ ratio.

The LQ model has been extended to fractionated radiotherapy with the derived biological effective dose (BED) widely used for comparing fractionation schedules and for calculating alternative regimes, for example to correct for an unwanted interruption of treatment. Extensions to the BED formulation have been developed to account for factors such as the incomplete repair of sub-lethal damage and repopulation which can occur during fractionated regimens (for more details, see the article on Biological effective dose).

The LQ model generally works well in describing the response to radiation both in vitro and in vivo for the doses typically used in treatment schedules utilising multiple daily fractions. However, the LQ formulation results in continuously bending cell survival curve which does not match experimental observation if survival curves are determined for very high doses. In such cases, the dose–response relationship approximates to a straight line in a log-linear plot, that is cell killing is an exponential function of dose. It has been proposed that a more kinetic model such as the lethal-potentially lethal (LPL) model of Curtis (1986) is more appropriate to describe the response with large fraction sizes such as those used in stereotactic radiosurgery. Guerrero and Allen Li (2004) have proposed a modified LQ model, an extension of the conventional LQ model, to more accurately describe high-dose regimes.

The LQ model does not adequately describe the cellular response to radiation at low doses, below about 1 Gy. It has been shown that many mammalian cell lines exhibit hypersensitivity below about 10 cGy, characterised by a cell survival curve slope considerably steeper than that expected by extrapolating back the response from high-dose measurements, Figure L.28. A review of the evidence and possible mechanisms for low-dose hypersensitivity can be found in the papers by Joiner et al. (2001) and Marples and Collis (2008). The LQ model has been modified to take account of this phenomenon resulting in the induced repair model, Equation L.2.

$$\text{Surviving fraction, } SF = \exp \left[ -\alpha D \left[ 1 + \left( \frac{\alpha}{\alpha_r} - 1 \right) \times \exp \left( -\frac{D}{D_c} \right) \right] - \beta D^2 \right] \quad (L.2)$$

- $D_c \sim 0.2 \text{ Gy}$
- At very high doses ($D \gg D_c$), equation $\rightarrow$ LQ model with parameters $\alpha$ and $\beta$
- At very low doses ($D \ll D_c$), equation $\rightarrow$ LQ model with parameters $\alpha$ and $\beta$

The induced repair model is a modification of the LQ model to account for the hyper-radiosensitivity observed at very low doses.

The LQ model also requires modification to account for the effect of dose rate on response. As stated earlier, the standard LQ

**FIGURE L.27** The LQ model provides a description of the continually downward bending form of the cell survival curve.

**FIGURE L.28** Many mammalian cells have been shown to exhibit hyper-radiosensitivity characterised by a considerably steeper slope, $\alpha_r$, than that expected by extrapolating back the response from high-dose measurements, $\alpha$. (Adapted from Joiner, M.C. et al., *Int. J. Radiat. Oncol. Biol. Phys.*, 49, 379, 2001.)
model generally works well for treatment schedules utilising multiple daily fractions at the dose rates used clinically for external beam radiotherapy. In such cases, dose delivery takes no more than a couple of minutes. However, as dose rate is lowered, the time taken to deliver the radiation dose is extended and it becomes possible for the radiation response to be modified as a result of the following processes: repair of sub-lethal damage, redistribution, repopulation and reoxygenation. Such considerations are required in LDR brachytherapy and further details on the effect of dose rate on response can be found in the article on Dose rate dependence. The most widely used modified LQ model in such situations is the incomplete repair model of Thames (1985) shown in Equation L.3:

\[ E = \alpha D + \beta D^2 g(t) \]  

\[(L.3)\]

where

\[ g(t) = \frac{1}{\mu} \left[ \mu - 1 + \exp(-\mu t) \right] \]

- \(E\) is the level of biological effect
- \(\alpha\) and \(\beta\) are the LQ model parameters
- \(g\) depends on half-time for recovery \((T_{1/2})\) and duration of exposure \((t)\) where

\[ \mu = \frac{0.693}{T_{1/2}} \]

The incomplete repair model is a modification of the LQ model to account for the effect of low dose rate on radiation response.

A comprehensive review of the various parameterisations of the LQ model used in radiotherapy can be found in the book by Dale and Jones (2007).

**Abbreviations:** BED = Biological effective dose, LDR = Low dose rate, LET = Linear energy transfer, LPL = Lethal-potentially lethal, LQ = Linear quadratic and SF = Surviving fraction.

**Related Articles:** Alpha beta ratio, Cell survival curve, Dose rate dependence, Dose response model, Linear energy transfer, Low dose rate (LDR), Radioisensitivity, Relative biological effectiveness (RBE), Redistribution, Repair, Repopulation, Stereotactic radiosurgery, Surviving fraction


**Linear-quadratic dose–response curve** *(Radiotherapy)* The linear-quadratic model is the dose–response model commonly used to describe the shape of cell survival curves. For further information see the article on Linear quadratic (LQ) model.

**Related Articles:** Alpha beta ratio, Cell survival curve, Dose response model, Linear quadratic (LQ) model, Surviving fraction

**Linear stopping power** *(Radiation Protection)* The energy lost from a beam of charged particle (e.g. alpha or beta) ionising radiation per unit distance travelled through a medium is known as the linear stopping power of the material traversed.

For more information, see Stopping power.

**Related Article:** Stopping power

**Linearly polarised (LP)** *(Magnetic Resonance)* If the B1 field vector (i.e. the magnetic component of the transmitted RF pulse) at a point in space points in a constant direction then the field is said to be linearly polarised.

A linearly polarised B1 field is physically equivalent to the summation of two circularly polarised fields propagating in the same direction but with opposite directions of rotation.

For example, the linearly polarised wave shown in Figure L.29 can be written as

\[ B = B_1 \cos \omega x \]

This can be decomposed into

\[ B = \frac{1}{2} B_1 (\cos \omega x + \sin \omega x) + \frac{1}{2} B_1 (\cos \omega x - \sin \omega x) \]

where the two terms represent vectors rotating with opposing sense.

In a linearly polarised RF excitation pulse only the circularly polarised component with the same sense as the direction of precession is effective in exciting spins.

**Liquid chromatograph** *(Radiation Protection)* Liquid chromatography is a chemical analytical technique based on the separation of a compound, for example radiopharmaceutical, in different components (molecules) when passing it through a layer (solid or gel) with the use of some solvent (liquid phase) as an eluent. This technique is similar to filtration but in this case the separation of the mixture is a consequence of molecular interactions among the components.
of a compound and the layer (membrane or column) producing different transit times, that is some molecules (components) pass through the layer faster than others. Gel chromatography can be used for separating proteins of different molecular weights, drugs, etc. or for detecting and separating impurities in radiopharmaceuticals.

There are three general kinds of liquid chromatography:

1. **TLC (thin layer chromatography):** A layer of 0.01–2.00 mm, for example silica gel is put on a glass or plastic plate. The separation results from a slow penetration of this layer by the solution under examination.

2. **Column chromatography:** A layer is introduced in the column and then the solution passes through it. As an example is HPLC (high-performance liquid chromatography) that is used for the purification and identification of chemical compounds, for example radiopharmaceuticals. In this technique the phase polarity is significant. The sample is injected to the column and then eluent is pumped under high pressure (higher than 100 atm). The concentration of the different components of the eluate that are separated by the column is measured by a detector. The detector response is proportional to the concentration that is plotted as a function of time. The detectors used can be a simple UV–VIS spectrophotometer (measures the absorbance), a fluorescence spectrometer, a mass spectrometer, a laser scattering spectrometer or a radiation detectors, for example NaI(Tl) for measuring radioactivity of every component for radioactive samples.

3. **Electrophoresis:** For example, CE (capillary electrophoresis) is like a thin layer chromatography but the separation is made by the application of an electric field. The sample is introduced to the inlet of a capillary and then the high voltage, for example 30 kV is switched on. This technique makes possible separation of anions and cations in organic and non-organic salts. The detector is placed at the outlet of a capillary.

**Abbreviations:** CE = Capillary electrophoresis, HPLC = High-performance liquid chromatography and TLC = Thin layer chromatography.

**Related Article:** Chemical exchange


**Liquid crystal display (LCD)**

(Diagnostic Radiology) A liquid crystal display or LCD is an electronic display that uses cells of liquid crystal molecules to create pixels. By varying the electric field across the liquid crystal cells the opacity of each cell can be changed. If the cell is backlit the change in opacity causes a change in the viewed brightness of the pixel.

Active matrix flat panel thin film transistor (TFT) LCDs are used widely in modern computing, and high resolution displays are used in diagnostic radiology for viewing clinical images. These displays use individual liquid crystal cells to form pixels, which are controlled by an active matrix.

The term liquid crystal is used to describe a substance in a state between liquid and crystal. It is a liquid in which the molecules exhibit long-range order, a periodic pattern of atomic positions that extends over many atoms. A nematic liquid crystal is a substance that has different physical and optical properties in different directions because of the spatial anisotropy (elongated molecules) of the molecules, which are aligned in a regular chain. This anisotropy of the molecules leads to birefringent properties whereby it has different refractive indices parallel and perpendicular to the molecules. Therefore, light, which is polarised in the direction of the long axis of the molecules is absorbed whilst light polarised along the short axis is not.

The simplest form of an LCD is constructed from a thin layer of an organic compound whose cylindrical molecules tend to line up parallel to each other. The organic compound, or liquid crystal (LC), is held between two parallel glass substrates which have had a thin, transparent, layer of metal oxide deposited upon them, which act as electrodes. On either side of the substrates is placed crossed polarisers, polarisers placed at 90° from each other. Thin scratches on the glass substrate align the molecules at each surface in the direction of polarisation, creating a twisted nematic (TN) array. The TN twists the plane of polarisation of light so it is able to pass through both crossed polarisers, see Figure L.30. To change the degree of twist within the LCD, electrodes are placed on the glass substrate, and when a potential difference is applied across the LC an electric field is created causing the molecules to untwist and become almost perpendicular to the substrates so that light can no longer pass through both polarisers, see Figure L.31.

To create a monochrome image an array of liquid crystal cells is created and each LC cell is backlit, meaning a light source is placed behind it. Each cell appears bright when no voltage is applied across it (Figure L.30); as the electric field increases the molecular array...
moves and proportion of light able to pass through both polarisers decreases until a maximum voltage is applied and the cell appears black, Figure L.31. By altering the voltage all greyscales of an image can be reproduced.

The TN LCDs were the first type of liquid crystal display to be developed and their use was restricted by low contrast ratio, narrow viewing angle and slow response time. To overcome these issues different manufacturers have developed LCDs based upon several different molecular alignment and electrode patterns. These are twisted nematic (TN), in-plane switching (IPS) and vertically aligned (VA) LCD.

Related Article: Active matrix liquid crystal flat panel display

Liquid flow counting

(Nuclear Medicine) Liquid flow counting can be used as an on-line monitoring of a radioactive solution flowing through a plastic tube, for example in experiment set-up with column chromatographic techniques. Different radiation detectors, such as Geiger–Müller, CdTe, NaI(Tl) may be used depending on the application and the decay characteristics of the radionuclide. The detector can be connected to a single- or multi-channel analyser.


Liquid metal bearing

(Diagnostic Radiology) One of the latest x-ray tubes with rotating anode use special spiral groove bearings using as lubricant liquid metal (eutectic alloy of Gallium, Indium and Tin) which has melting point of −10°C. At room temperatures this alloy is liquid as mercury, but has lower vapour pressure. This tube is patented by PHILIPS®.

The spiral groove bearings (Figure L.32) have special profile (as in the car tires), which allows very smooth rotation over the ‘liquid metal’, using an effect similar to the familiar ‘aqua-planning effect’. This allows for a very low friction (i.e. excellent rotation) and maximal heat dissipation through the liquid metal. The minimal friction allows the motor to be rotated at operational speed at the beginning of the day and then run continuously (until stopped with electromechanical brakes).

Related Articles: Glass envelope, Bearing, Filament heating, X-ray tube, Anode, Metal x-ray tube

Liquid scintillation (LS) counting

(Radiation Protection) Liquid scintillation counting technique is used to measure weakly penetrating radiation like gamma radiation of energy less than 20 keV, β-rays of energy 15–2000 keV and α-emitters.

The counting efficiency \( \epsilon_{\text{count}} \) defined as

\[
\epsilon_{\text{count}} = \frac{\text{number of pulses recorded}}{\text{number of radiation quanta incident on detector}}
\]

is equal ≈100%.

The radioactive sample is mixed with liquid scintillator, placed in a sample holder (vial) transparent to optical radiation and then placed between two PM tubes (Figure L.33). The pulses registered by PM1 and PM2 tubes pass to a coincidence system. The coincidence system allows passing to an amplifier only pulses registered simultaneously by PM1 and PM2 (deriving from the scintillation of the same molecule). The PHA analyses its height (amplitude) and the result is presented in the recording device in analogue or digital form. The LS equipment is calibrated with an isotope with known energy of particles and in this way it is possible to identify measured radioisotopes and its quantity in the sample.

The detection output strongly depends on counting quenching. Quenching can be caused by interference in optical photons energy transfer by the sample and liquid scintillator (chemical quenching), by dilution of the sample in scintillation solution (dilution quenching) or by coloured substance (colour quenching). There are special experimental methods to take the quenching into account and to obtain accurate counting of measured sample. Chemoluminescence can also be a problem.

The β- radioisotopes often detected by LS counting include: H-3, C-14, P-32, S-35.

**Abbreviations:** LS = Liquid scintillation, PHA = Pulse height analyser and PM = Photomultiplier.

Related Articles: Chemical quenching, Coincidence circuit for liquid scintillation counters, Dilution quenching, Pulse height analysers, Scintillator


**Figure L.32** Spiral grooves on the rotor of an x-ray tube (the sectioned large anode is seen on the right of this model). Image taken with permission from a Toshiba model of power x-ray tube with liquid metal anode bearing. (Image courtesy of Toshiba, Shizuoka-ken, Japan.)

**Figure L.33** Scheme of a liquid scintillation counting equipment.
Lithotripter
(Ultrasound) A lithotripter is a device which is used for non-invasive treatment of kidney stones. It works by disintegrating the stones by extracorporally induced ultrasonic shock waves. The shock waves form due to non-linear propagation and can have peak positive pressures in the range of 100 MPa.

The German company Dornier Systems originally developed the method. Today the shock wave can be generated in basically two ways, either by a point source or by an extended source. The most frequently used point source is a spark gap, placed in one of the foci of an ellipsoid, immersed in water. As a high voltage is discharged over the gap, an explosive plasma formation and vaporisation of the water takes place. As all rays from the first foci, via the perimeter of the ellipsoid, to the second foci have equal path length, the energy will again converge in the second foci, placed where the stone is located. For an extended source, a large number of ultrasound transducers (many hundreds) are located in an array in the shape of a segment of a sphere. Simultaneous transmission of a high amplitude pulse on all elements will also produce a shock wave towards the centre of the sphere. A treatment using either strategy, can last up to an hour, during which up to 10,000 shock wave pulses are emitted. The stone is slowly disintegrated, and not pulverised by one single blow. The underlying mechanisms are not fully understood, but the high pressure and rise-time are believed to be of importance, as well as cavitation effects.

Local area network (LAN)
(General) A LAN is a small to medium scale computer network (as opposed to wide area networks WANs) providing information access and data sharing within a home or small office setting. The LAN can link up and offer a communication route between computers acting as Servers or just Workstations, printers, network attached storage systems and other networkable devices.

The LAN usually offers much higher data transfer rates than are available over WANs although this is dependent on the exact architecture and physical medium used to relay the data. The LAN may take many physical forms including Ethernet over coaxial cable or shielded or unshielded twisted pair or Wi-Fi and can be configured in several organisational forms or topologies. Switches, routers or hubs can be added to a LAN to transfer information packets between separate sections of the LAN and link the LAN to one or more other LANs or WANs usually via a modem.

Local overdose
(Radiotherapy) A local overdose is delivered to a patient when a higher dose is given than was intended; this can arise from accidental equipment malfunction or a miscalculation in the treatment plan or the applied monitor units. It is generally accepted that the radiation dose should be delivered to a tolerance of within 5% of the prescribed dose with a confidence level of 95%, therefore a level of 10% as a definition of overdose would seem reasonable. However this may vary across different quality assurance systems. An overdose may increase the frequency and severity of treatment complications. Complications depend on several factors such as the total delivered dose, the total duration of the treatment, the size and location of the irradiated volume and the organisation of the functional subunits of the organ being irradiated.

Local rules
(Radiation Protection) In the work with ionising radiation, employers, registrants and licensees shall, in consultation with workers, through their representatives (if appropriate): (1) ensure protection and safety for workers and other persons; (2) comply with regulations and dose limits, including investigation levels or authorised levels and procedures in the event that any such dose limit is exceeded and (3) ensure that any work be adequately supervised. In order to achieve safety in the procedures it is important to prepare the local rules and made them known to workers and any other person involved.

The local rules should be tailored taking into account the local laws and regulations, for the various medical applications of ionising radiation and refer, when possible, to specific equipment and procedure; they should include: procedures for wearing, handling and storing personal dosimeters as well as actions to minimise radiation exposure during unusual events.

The workers should be well informed and trained on the content of the local rules. In addition a written copy should be readily available in each room, where eventually could be needed. The persons responsible with their telephone number should be clearly indicated in case help is needed.


Local underdose
(Radiotherapy) A local underdose consists of delivering less than the intended dose to a patient because of accidental equipment malfunction or a miscalculation in the treatment plan or applied monitor units. Underdosing may reduce the tumour control probability. Many underdoses go undiscovered and may only be detected after a relatively long time and consequently may involve a large number of patients.

Localisation jig
(Radiotherapy) A localisation jig is the device with embedded fiducial markers used in brachytherapy for source localisation. Source localisation is the determination of the 3D coordinates and orientation of each source relative to the patient anatomy. Some sources refer to the localisation jig as ‘reconstruction jig’.

The jigs are made of acrylic or plastic and contain fiducial markers in the shape of crosshairs with known sizes on each side of the jig. The crosshairs are used by the planning software to create
Localisation radiograph

(a common coordinate system. For the geometry reconstruction, two appropriate projection images are needed. A single point in space (3D) results in two coordinates in each projection image (2D). Known geometry of the localisation jig and the fiducial markers enables the determination of coordinates for other structures projected on images (Figure L.34).

Related Articles: Brachytherapy, Afterloading


Localisation radiograph

(Radiotherapy) Localisation is the process of determining the tumour/target location. It is the first stage in the radiotherapy treatment planning process. This process may be achieved using tomographic imaging, such as CT or MRI or by using projection radiographs. Localisation is often achieved by using two orthogonal radiographs to determine the position of the tumour or an anatomical structure in three dimensions.

Treatment Simulator: A treatment simulator may be used to obtain a localisation radiograph. This is a diagnostic x-ray set mounted on an isocentric gantry such that it has the same degree of movement as a treatment linac. The localisation radiograph may be obtained with contrast medium to locate the tumour in certain sites such as the bladder. Figure L.35 shows a treatment simulator.

Abbreviations: CT = Computed tomography and MR = Magnetic resonance imaging.

Localiser

(Magnetic Resonance) The term localiser normally refers to the acquisition of three orthogonal slices of a volume of interest in order to facilitate pre-examination slice positioning. A localiser scan commonly employs a $T_1$-weighted gradient-echo pulse sequence and is also referred to as ‘Scout’ or ‘Survey’ depending on the MRI unit vendor.

Related Articles: Gradient echo (GE), $T_1$-weighted

Logic analyser

(General) A logic analyser is an electronic instrument that measures and displays signals in digital circuits. Logic analysers are used for capturing data in systems that have many digital channels and therefore cannot be displayed with an oscilloscope. It usually has an embedded computer which runs software enabling conversion of the measured data into a form more appropriate for analysis: timing diagrams, assembly language, state machine traces, or others (Figure L.36).

Hyperlink: http://en.wikipedia.org/wiki/Logic_analyzer

Longitudinal magnetisation

(Magnetic Resonance) The magnetisation of tissues in the direction of the static magnetic field $B_0$ is designated as longitudinal magnetisation. If a tissue is placed inside a magnetic field $B_0$ the magnetic moments of individual protons (=spin) will begin to rotate, or precess, about the magnetic field. The spins will be tilted slightly away from the axis of the magnetic field, but the axis of rotation will be parallel to $B_0$. The tissue will therefore become magnetised in the presence of $B_0$ with a value $M_0$ known as the net magnetisation (i.e. vector sum of all spins). Longitudinal magnetisation cannot, however, directly produce a RF signal. If a radiofrequency pulse is applied along the $x$-axis it...
bends the magnetisation $M_0$ away from the z-axis and causes it to precess around a new direction ($B_1$). A RF pulse, indicated as $90^\circ$ pulse and creating an additional magnetic field $B_1$, with a central frequency $f_0$ and orientation perpendicular to $B_0$, resulting in an effective field $B_{eff} = B_0 + B_1$ will cause $M_0$ rotate entirely into the transverse plane (Figure L.37).

There will be no longitudinal magnetisation following the $90^\circ$ pulse. The longitudinal magnetisation along the z-axis has been converted into a detectable magnetisation in the 90° pulse. The longitudinal magnetisation along the transverse plane (Figure L.37).

This return of magnetization follows an exponential growth process with $T_1$ being the time constant for the grow (Figure L.38). After three $T_1$ time periods $M$ will return to 95% of its value prior to the excitation.

**Longitudinal movement**  
(Nuclear Medicine) This is movement along the long axis (feet to head) of the body. Patients in scanners or emission cameras can induce image artefacts originating from patient movement during acquisition. Motion artefacts can be difficult to correct for and may lead to misdiagnosis if not properly attended to. It is therefore important to minimise patient movement during imaging.

**Longitudinal travel**  
(Nuclear Medicine) See Longitudinal movement

**Longitudinal wave**  
(Ultrasound) When an ultrasound wave propagates in a medium, the particles within the medium will start oscillating. In gases, liquids and soft tissue this oscillation is always in the same direction as the ultrasound wave, giving rise to longitudinal (or compressional) wave propagation, Figure L.39. The particles’ displacement and velocity depend on the medium’s acoustic impedance, the pressure amplitude and the frequency of the ultrasound wave (see Displacement).

**Related Articles:** Displacement, Compression, Transversal wave, Lamb wave

**Lookup table**  
(Diagnostic Radiology) Lookup table (look-up table, LUT) is an array of data used by the computer to convert data from one form to another. LUT are widely used in image processing, for example for storing palettes of colour (e.g. the 8 bits in a byte used to address 256 cells each holding information about a particular colour). A typical use of LUT in medical imaging is for the conversion of the pixel numbers into physical colours (or shades of grey).

**Lorentzian lineshape**  
(Nuclear Medicine) The Lorentzian lineshape refers to shape of a Lorentzian distribution, also known as a Cauchy distribution. The Lorentzian distribution has a similar bell shape as the normal distribution. The probability density function for the Cauchy distribution is

$$P_{Cauchy}(x; x_0, \gamma) = \frac{1}{\pi} \frac{\gamma}{(x - x_0)^2 + \gamma^2}$$

where $x_0$ is the mean value (the centre of the distribution) and $\gamma$ is the full width at half maximum.

**Hyperlink:** [http://mathworld.wolfram.com/CauchyDistribution.html](http://mathworld.wolfram.com/CauchyDistribution.html)

**Lossless compression**  
(General) Image compression can either be lossless or lossy, depending on whether the image can be subsequently reproduced exactly (lossless), or whether data are lost in the compression (lossy) (Figure L.40).

Lossless images undergo no quantisation, which is the main source of loss for lossy image files. The encoding carried out for lossless files is based on redundancy reduction schemes, which reduce the overall image size but maintains all the information present.

**Redundancy Reduction Schemes:** Firstly the images are treated for interpixel redundancy, taking advantage of any patterns present. One example of this is run length encoding (RLE)
in which consecutive equal pixel values are replaced with a ‘value-runlength’ couplet. For example, 20 px of value 7, can be instead replaced by the couplet 7–20, which is a reduction of 90% in required storage space. This algorithm works especially well for greyscale images and overall achieves compression ratios of around 2:1.

The second stage reduces the coding redundancy present. For example, Huffman coding assigns short codes to the most frequent values, only using longer codes for those values that occur rarely. Although the coding table needs to be saved also, this is usually a small overhead.

**Lossless File Formats:** An example of lossless file format is the graphics interchange format (GIF), which allows images to be faithfully restored after compression as no information is thrown away. It does have the disadvantage that images with greater than 8-bit colour resolution need to be reduced in resolution before they can be compressed. However this has been addressed in the newer portable network graphics (PNG) file format, which is becoming the scheme of choice for lossless image compression.

Lossless file formats should be the first choice for medical applications, where there is the risk of a mis-diagnosis from distortions or information lost through a lossy compression process, or even where the diagnostic value is yet unknown.

**Abbreviations:** GIF = Graphics interchange format, PNG = Portable network graphics and RLE = Run length encoding.

**Related Article:** Lossy

**Lossy compression**

(General) Image compression can either be lossless or lossy, depending on whether the image can be subsequently reproduced exactly, or whether data are lost in the compression.

Lossy compression is more efficient, but the extent of compression will depend on the tolerance of the observer for degradation of the image. An example of a lossy file format is JPEG (joint photographic experts group), which allows the user to set a ‘quality factor’ which specifies the amount of data thrown away. At low-quality factors, JPEG artefacts may appear such as the appearance of ‘blockiness’ within the image. The transformation step for JPEG makes use of the discrete cosine transform, to shift the colour data into a more suitable mode for compression and encoding. The next quantisation step (see Figure L.40) is the main source of loss, where the colour values are simplified according to the quality factor. Finally the data are encoded using similar methods to the lossless compression, using both run length encoding and Huffman coding.

Lossy compression is often the compression algorithm of choice for photographs, allowing a useful trade off between quality and size/time. However line drawings or graphs do not lend themselves to JPEG, where any sharp changes can be distorted. It is recommended that images in a master archive are not stored in a lossy format, and it must be noted that frequent retrieval, modification and archiving will probably degrade the image further each time. Hence lossless file formats should be the first choice for medical applications, where there is the risk of a misdiagnosis from distortions or information lost through the compression process, or where the diagnostic value of the images may not be known fully.

**Abbreviations:** DCT = Discrete cosine transform and JPEG = Joint photographic experts group.

**Related Article:** Lossless compression

**Low contrast**

(Diagnostic Radiology) There are several terms specifying the subject contrast (mainly used in x-ray radiography). The most often used ones are high contrast and low contrast.

Low contrast is used to describe images with small differences between the optical densities of the adjacent objects. Low contrast detectability is largely influence by the noise in the image. A number of synonyms are used in practice – as soft contrast or long scale contrast. The latter refers to the fact that x-ray films with big latitude present a large (long) scale of gray levels. For more detail see the article on High contrast.

**Related Articles:** Subject contrast, High contrast, Film latitude

**Low dose rate (LDR)**

(Radiotherapy, Brachytherapy)

**Dose Rates in Brachytherapy:** Different dose rates are used in brachytherapy, connected to different treatment techniques. ICRU, the International Commission on Radiation Units and Measurements, defined these dose rates in its Report No. 38 ‘Dose and Volume Specification for Reporting Intracavitary Therapy in Gynaecology’:

1. Low dose rate, LDR
   a. 0.4–2.0 Gy/h
   b. Traditional radium technique; 0.5 Gy/h, 60 Gy with treatment time 5 days
   c. Large amount of clinical data
   d. (NOTE! Ultra low dose rate 0.01–0.3 Gy/h!)
2. Medium dose rate, MDR
   a. 2–12 Gy/h
   b. More seldom used
3. High dose rate, HDR
   a. >12 Gy/h = 0.2 Gy/min
   b. Treatment times approx. 5–20 min (external beam therapy belongs here)
   c. Clinical data available
4. Pulsed dose rate, PDR
   a. Mimics LDR, using many small ‘HDR pulses’ during a longer treatment time Example: 1 pulse per hour during 24 h, 0.5 Gy per pulse given in 5 min; total dose 12 Gy/day
The radiobiological effects in the tissues irradiated depend on the type of applicator used, on the fractionation scheme and on both dose and dose rate distributions. As stated in the ICRU Report 38: ‘the clinical experience accumulated with radium techniques cannot be applied to new irradiation conditions without careful consideration’. This includes consideration of both tumour effects and effects on normal tissues.

**Abbreviation:** ICRU = International Commission on Radiation Units and Measurements.

**Related Articles:** Brachytherapy, Dose rates in brachytherapy, see also articles under radiobiology

**Further Reading:** ICRU (International Commission on Radiation Units & Measurements, Inc.). 1985. Dose and volume specification for reporting intracavitary therapy in gynecology. ICRU Report 38, Bethesda, MD.

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**Low melting point alloy**  
*(Radiotherapy)* Frequently part of the radiation field has to be shielded to avoid irradiating underlying sensitive structures. This can be done using alloys with a low melting point which can be poured in molten form into a polystyrene mould and then allowed to harden at room temperature. Although in many centres they are largely confined to use with electrons, they are especially useful when complex field shapes are required and blocks can be shaped to account for beam divergence. The alloy can also be melted down and reused. The use of these materials has replaced lead in many cases which often needed to be milled if complex shapes were required and had a high melting point (327.5°C) making pouring and moulding much more difficult.

Popular low melting point products are Cerrobend® and Ostalloy® which are alloys of lead, cadmium, bismuth and tin. These have a melting point of 69°C and 70°C, respectively, although they are usually worked at temperatures of 90°C and 95°C, respectively, since they are easier to pour and mould at these temperatures. Newer materials such as MCP 96®, melting point 96°C, is an alloy of lead, bismuth and tin with no cadmium content, the latter having a relatively high toxicity.

Multi-leaf collimators (MLCs) have largely replaced the use of blocks from low melting point alloys and lead for shielding of megavoltage photon beams but sometimes blocks are required when adequate shielding cannot be achieved with MLCs. This would be the case when the MLC shielding profile is not smooth enough for the task because of the leaf width or when a shielded island within a field is needed.

Centres using low melting point alloys for electron treatment normally use a standard thickness of shielding of around 15 mm. This is because for electron energies from 4 to 18 MeV a typical thickness of lead shielding would have been 10 mm. Twenty per cent more alloy is needed for lead equivalence so 12 mm would be required and in practice 15 mm is used, the additional 3 mm is added as a precaution against small air bubbles forming in the alloy as it sets.

Output factors must also be measured for these irregular fields shaped by cut-outs by comparing the output with standard open field.

**Abbreviation:** MLCs = Multi-leaf collimators.

**Related Articles:** Block design, Block tray, Custom blocking, Cerrobend, Block transmission factor, Output factor


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**Low-pass filter**  
*(General)* Image processing of medical images often involves some kind of low-pass filtering to reduce the noise in the image that otherwise can be disturbing to the reader of the image. In the Fourier domain signals is represented by amplitude and frequency. In Fourier-transformed images, the frequency has the unit of cm⁻¹. This means that small details are represented by high frequencies and large structures are represented by low-frequencies. A Fourier spectrum describes a signal in terms of amplitudes and frequencies. The noise is usually disturbing the image as the type of mottle or ‘salt-and-pepper’ type, that is local variation in pixel values that does not have any physiological meaning. This noise will appear in the high-frequency part of the spectrum. It is therefore the purpose of a low-pass filter to reduce these high-frequency amplitudes and let the low-frequency amplitudes pass through the filtering step. A mean-value calculation of a number of samples is also a low-pass filter that reduces the local variations. Mean-value filters or weighted mean-value filters are therefore common in nuclear medicine work stations. A commonly used 2D filter is the Butterworth filter that is described as

\[
T(u,v) = \frac{1}{1 + \left( \frac{D(u,v)/D_0}{2} \right)^n}
\]

The filter has two parameters, namely, the order \( n \) and the cut-off frequency \( D_0 \). The order controls the slope of the descent and the cut-off value defines when the descent will start. Note that the cut-off frequency is the frequency at which the amplitude has been reduced to 0.5.

**LSF**

See Line spread function (LSF)

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**L-shell**  
*(General)* The discovery that electrons in atoms occupy different energy shells was made by Barkla and Moseley from studies of x-ray absorption in atoms. These shells were originally labelled by Barkla with the letters K, L, M, Q. At present there are two equivalent nomenclatures used to describe these energy shells: quantum numbers and the x-ray notation of Barkla.

**Quantum Numbers:** Four quantum numbers are required to specify the state of an electron in an atom:

- The principal quantum number \( n \) specifies the electron energy, it takes values of \( n = 1, 2, 3, \ldots \)
- The angular momentum quantum number \( \ell \) specifies the magnitude of the orbital angular momentum; \( \ell \) takes values from 0 to \( n - 1 \). It should be noted that the following notation is applied:

\[
\ell
\]

\[
0 \quad s
\]

\[
1 \quad p
\]

\[
2 \quad d
\]

\[
3 \quad f
\]

\[
4 \quad g
\]

\[
5 \quad h
\]

- The magnetic quantum number \( m_\ell \) specifies the orientation of the orbital angular momentum, it takes values from \( -\ell \) to \( +\ell \).
- The spin angular momentum \( m_s \) specifies the spin angular momentum of the electron, that is spin up (+1/2) or spin down (−1/2).
For example, the electronic structure of carbon, which has six electrons, is often written as 1s\(^2\) 2s\(^2\) 2p\(^2\); this implies that it has three occupied orbitals, the two most tightly bound electrons are in the 1s orbital, and there are a further two electrons in both the 2s and 2p orbitals.

**X-Ray Notation:** A second way of labelling electron orbitals is to use x-ray notation, a convention commonly employed in x-ray spectroscopy. In this notation the principal quantum number (\(n\)) is attributed a letter as shown in the following:

<table>
<thead>
<tr>
<th>Quantum Number Notation</th>
<th>X-Ray Notation</th>
</tr>
</thead>
<tbody>
<tr>
<td>1s</td>
<td>K(_1)</td>
</tr>
<tr>
<td>2s</td>
<td>L(_1)</td>
</tr>
<tr>
<td>2p ((m_s = +1/2))</td>
<td>L(_2)</td>
</tr>
<tr>
<td>2p ((m_s = -1/2))</td>
<td>L(_3)</td>
</tr>
<tr>
<td>3s</td>
<td>M(_1)</td>
</tr>
</tbody>
</table>

Hence, there are three L-shells: L\(_1\), L\(_2\) and L\(_3\). Some related terms relevant to medical physics include the following:

- The L-edge which is the energy of x-rays at which the L-orbital begins to absorb.
- To describe an Auger peak three orbitals are needed, firstly the core hole level, secondly the relaxing electron’s initial state and thirdly the emitted electron’s initial state, for example KL\(_1\)L\(_2\).

**Related Articles:** Atom, Auger electron, Electron

**Lubberts’ effect** *(Diagnostic Radiology)* Lubberts’ effect (named after G. Lubberts, who first published on the effect in 1968) is related to the use of phosphor materials in x-ray imaging. If x-rays are absorbed in the upper layer of the phosphorus the light produced at the absorption point travels through the thickness of the phosphorus, thus it spreads and leads to image blur. The effect is important for thick phosphorus layers, as in image intensifier, or in indirect flat panel digital detectors using phosphorus. In the latter if the light is absorbed in the lower layer of the phosphorus, it reaches immediately the detector and the image is with less blur. The Lubberts effect is related with the MTF and noise in the image and results in decrease of the DQE (detective quantum efficiency) with the increase of the spatial frequency.

**Related Articles:** Detective quantum efficiency, Screen selection


**Luminance** *(Diagnostics Radiology)* See HSL (hue, saturation luminance)

**LUT** *(Diagnostics Radiology)* See Lookup table
**M-mode**

*(Ultrasound)* M-mode, or motion mode, is based on a one line
B-mode measurement, which is continuously repeated (Figure
M.1). The penetration depth is shown on the y-axis, and the x-axis
shows time (Figure M.2). Stationary reflecting targets are shown as
straight lines and moving targets are shown as a repetitive pattern.
M-mode is very useful when the degree and rate of motion is to be
measured and is used, for example, for investigation of heart valves.

*Related Articles:* A-mode, B-mode

**mA selector**

*(Diagnostic Radiology)* The mA selector is part of the filament
circuit controls of the high voltage generator (HVG) of an x-ray
equipment. In classical HVG it selects the filament current through
a variable resistor connected at the primary side of the filament
transformer. Additionally this circuit includes a set of filament
resistors for various combinations (low kV at high mA; high kV at
low mA; low kV at low mA).

Contemporary equipment with high frequency generator use
a system which changes the frequency supplying the filament
transformer, thus varying the filament current (hence the tube
current, mA).

Direct measurement of the mA is possible only in the second-
ary circuit of the high voltage transformer (in the middle point of
the secondary winding) – see the block diagram H.25 in the article
*High-voltage generator*. During quality control one normally mea-
sures the mA variation with kV (mainly the effect of voltage drop).
In this measurement errors above 10% are considered unaccept-
able. Output dose measurement at constant kV can also be an indi-
cator for the mA change, but can not supply accurate information
for the mA.

*Related Articles:* High voltage generator, High frequency
generator, Voltage drop, Filament circuit, Filament resistor

**Macroradiography**

*(Diagnostic Radiology)* Macroradiography is a radiographic
method producing enlarged x-ray images of the object, by placing
it at some distance from the detector/film. The method has advan-
tages in cranial angiography, mammography, etc., but is rarely used
these days. Macroradiography requires x-ray tubes with very small
focal spot (below 0.3 mm), as otherwise the enlarged image will be
blurred. If the object is placed mid-way between the focal spot and
the film (i.e. ×2 magnification), the size of the image blur (geomet-
ric unsharpness) will be equal to the size of the effective focal spot.
In this case, if 0.2 mm blur is accepted for this particular image,
the focal spot size should be 0.2 mm (i.e. the image of the object is
doubled, but the blur is still 0.2 mm). Macroradiography is different
(better) than simple optical enlargement, because when magnify-
ning glass is used, all image features (the object and its blur) will be
equally enlarged.

*Related Articles:* Focal spot, Geometric unsharpness,
Magnification, Object-film distance, Ultra-fine focus

**Magic angle**

*(Magnetic Resonance)* Solid materials feature a rigid arrangement
of spins. These constantly experience the local fields of each other
leading to inhomogeneous dipole–dipole broadening. From a mag-
netic moment \( \mu \) at a distance \( r \) and polar angle \( \phi \) a second spin
experiences a

\[
B_{loc} = \pm \frac{\mu_i}{r^3} (3 \cos^2 \phi - 1)
\]

as derived from the dipolar Hamiltonian. The ± signs reflect the
spin alignment with respect the applied field \( B_0 \). The solution of
the equation \( 3 \cos^2 \phi = 1 \) gives \( \phi = 1/3 \) or \( \cos \phi = \sqrt{1/3} \)
that is 55°44′ and called `magic angle`. By magic angle spinning
will be exposed to the presence of through-bond coupling, some of the A nuclei composed of two peaks (known as ‘singlets’ in this context). In spin ½. Neglecting scalar coupling, one would expect a spectrum of resonance peaks in the NMR spectrum of the molecule into field of the other. The fact that each nucleus can exist in more than nonzero spin are connected to each other via chemical bonds, so spins within a molecule. This occurs when two or more nuclei with nuclear dipolar interaction, chemical shift anisotropy effects and quadrupole field gradient interaction. Dipole–dipole interaction also affects MRI contrast in highly ordered structures like a tendon. At the magic angle, the $T_2$ of the tendon is slightly increased because collagen, which is responsible for the majority of tendon composition, has an anisotropic structure. This increase is negligible when TE is long. However, when TE is short as in $T_1$ or PD weighted images, the result will be an increased signal intensity.

Related Articles: Chemical shift, Dipole

Magnet
(Magnetic Resonance) A magnet is a material or object that produces a magnetic field. Magnets are divided into two groups, the permanent magnet and impermanent magnets like electro-magnets and superconductive magnets.

In all materials there are electrons that rotate in a specific path. The rotation generates a small magnetic field and if the movement of the electrons is coordinated, for example, by an outer magnetic field, they will amplify the magnetic field so it can be noticed outside the material. If the generated magnetic field stays on after the outer field is removed it is called a ‘hard’ or permanent magnet and if it disappeared with the outer field it is called ‘soft’. In nature there are some rocks, for example loadstone, that are hard magnets but if it disappeared with the outer field it is called ‘soft’. In nature there are some rocks, for example loadstone, that are hard magnets but if it disappeared with the outer field it is called ‘hard’ or permanent magnet and if it disappeared with the outer field it is called ‘soft’. In nature there are some rocks, for example loadstone, that are hard magnets but if it disappeared with the outer field it is called ‘hard’ or permanent magnet and if it disappeared with the outer field it is called ‘soft’. In nature there are some rocks, for example loadstone, that are hard magnets but if it disappeared with the outer field it is called ‘hard’ or permanent magnet and if it disappeared with the outer field it is called ‘soft’.

A hard magnet’s natural magnetism decreases when heated and increases when it is cooled.

Related Articles: Electro-magnet, Permanent magnet, Resistive magnet, Superconductive magnets

Magnet, permanent
(Magnetic Resonance) See Permanent magnet

Magnet, superconducting
(Magnetic Resonance) See Superconducting magnet

Magnetic coupling
(Magnetic Resonance) The term ‘magnetic coupling’ normally refers to what is more correctly termed ‘scalar coupling’ between spins within a molecule. This occurs when two or more nuclei with nonzero spin are connected to each other via chemical bonds, so that the spin state of a given nucleus influences the local magnetic field of the other. The fact that each nucleus can exist in more than one spin state ($\pm \frac{1}{2}$ in the case of a hydrogen nucleus) results in splitting of resonance peaks in the NMR spectrum of the molecule into ‘multiplets’ made up of several components.

It is helpful to consider the simple example of a molecule containing two interacting nuclear spins, A and X, each with spin $\frac{1}{2}$. Neglecting scalar coupling, one would expect a spectrum composed of two peaks (known as ‘singlets’ in this context). In the presence of through-bond coupling, some of the A nuclei will be exposed to X nuclei with spin state $+\frac{1}{2}$, and others to X nuclei with spin state $-\frac{1}{2}$. These two populations of A nuclei will experience slightly different local magnetic fields, so the spectral peak due to A nuclei will be split into two equal components. Similarly, the peak due to X spins will be divided into two due to the presence of two populations of A spins. The frequency differences between the components in each case are given by a coupling constant, J, typically 1–15 Hz for coupling between protons and up to 100 Hz or so if other nuclei are involved. J is independent of static magnetic field strength (Figure M.3).

If a larger number of nuclei are present, a triplet or higher order multiplet may arise, depending on the structure of the molecule and the nature of the coupled spins. The impact of coupling depends on the magnitude of J relative to the chemical shift, $\delta$, between the nuclei. The behaviour of spins in the strong coupling regime ($J \gg \delta$) is much more complicated than that of weakly coupled spins ($J \ll \delta$), and requires quantum-mechanical treatment. In the notation used to describe coupled spin systems, strength of coupling may be inferred from the alphabetical proximity of the letters used to designate the spins – for example, AX is a weakly coupled two-spin system and AB a strongly coupled one. The same principles extend to systems with more than two spins, such as $AX_2$, (e.g. lactate) and AMNPQ (e.g. glutamate). The strength of coupling may also depend on static magnetic field strength of the spectrometer used and weakens at higher fields, improving spectral assignment. In addition to SNR gains this points to using higher fields in MR spectroscopy.

Related Articles: Chemical shift, Decoupling


Magnetic dipole
(Magnetic Resonance) A magnetic dipole is a pair of magnetic poles of equal magnitude but opposite polarity separated by some small distance. Magnetic dipoles can be characterised by the dipole moment equal to the product of the magnetic strength of one of the poles and the distance separating the two poles. The direction of the dipole moment corresponds to the direction from the south to the north pole. Magnetic dipoles are created by current loops or by quantum-mechanical spin.

The strength of a dipole magnetic field is given by

$$B(r, \lambda) = (M/r^3)(1 + 3\sin^2 \lambda)^{1/2}$$

where

- $r$ is the distance from the centre
- $\lambda$ is the magnetic latitude ($90 - \theta$) and $\theta$ = magnetic colatitude
- $M$ is the dipole moment

Magnetic dipole moment
(Magnetic Resonance) The magnetic dipole moment $\vec{\mu}$ is the most elementary measure of the strength of a magnetic source and it constitutes an intrinsic property of fundamental particles (quarks

\[ (a) \]

\[ (b) \]

FIGURE M.3 Spectrum of two spin $\frac{1}{2}$ nuclei (a) in the absence and (b) in the presence of weak spin coupling.
and charged leptons). The magnetic dipole moment may be visualised by imagining the spin of the electric particles as a spinning gyroscope representing a tiny loop of electric current around the spinning axis. This current loop produces its own magnetic field. The connection between the magnetic dipole moment of the particle and the spin angular momentum vector \( J \) is
\[
\vec{\mu} = \gamma J
\]
where \( \gamma \) is a particle-specific constant called the gyromagnetic ratio.

**Related Articles:** Gyromagnetic ratio, Magnet


**Magnetic field lines** (Magnetic Resonance) The stray magnetic field outside the bore of the magnet is known as the stray fringe field. When a magnetic field is generated in a medium the response of the medium is its magnetic induction \( B \) also called flux density. A ferromagnetic object aligns itself along the direction of the magnetic flux, which is caused by the presence of a magnetic field in the medium. The magnetic field lines are a geometrical abstraction, which are used to visualise the direction and strength of a magnetic field. The direction of the magnetic field lines can be examined by using a magnetic dipole. The static magnetic field has no respect for the confines of conventional floors or ceilings. Knowledge of the extension of the magnetic field lines in the space permits determination of the area around the MR scanner in which specific safety rules must be followed. The MR site could be conceptually divided into four zones taking into account the extension and the value of the magnetic field isocountours (i.e. lines of equal magnetic field strength) Zone I includes all freely accessible areas. Zone II is the interface between the freely accessible Zone I and the strictly controlled Zones III and IV. It is in Zone II that the patient screening is performed as well as the screening of all the persons who have access to Zone III and IV. Zone III is the region in which free access of unscreened persons or ferromagnetic objects or equipment can result in serious injury or death as a result of interactions between the individuals or equipment and the MR static magnetic field. Zone III regions should be physically restricted from general public access by suitable physical locks. Zone III may project through floors and ceilings of MRI site, imposing magnetic field hazards on persons on floors other than that of the MR scanner. Zone IV, which is always located within Zone III, is the MR magnet room. The knowledge of the value and extension of the isocontours permits identification of the area within which the monitoring equipment might not properly function.

**Magnetic flux** (Magnetic Resonance) The magnetic flux through a surface area is the integral of the normal component of the magnetic field times \( \mu \) over the area:
\[
\psi_m = \iint \mu H \cdot ds
\]
where \( \mu \) is the permeability of medium (Henry/metre, H/m).

The SI unit of magnetic flux \( \psi_m \) is the weber (Wb). Dimensionally \( \psi_m = [ML^2IT^{-2}] \). The flux is conventionally represented by imagining lines of induction to be spaced so that the number through a given area is equal to the flux through that area.

**Magnetic flux density** (Magnetic Resonance) The magnetic flux density (SI unit Tesla) is one of the two magnetic vector fields in Maxwell’s equation.

The magnetic flux through a surface area is the integral of the normal component of the magnetic field times \( \mu \) over the area:
\[
\psi_m = \iint \mu H \cdot ds
\]
where \( \mu \) is the permeability of medium (H/m). The SI unit of magnetic flux \( \psi_m \) is the weber (Wb).

Dividing the magnetic flux by the area \( A \) gives the magnetic flux density \( B \) of flux per unit area. Thus
\[
B = \frac{\psi_m}{A} = \mu H
\]
The SI unit of magnetic flux density is Wb/m\(^2\) or Tesla, T.

The magnetic flux density \( B \) has the same direction as \( H \) in isotropic media with a magnitude \( \mu H \) or
\[
B = \mu H = \mu_r \mu_0 H \quad (\text{Wb/m}^2 \text{ or T})
\]
where
\[
B \text{ is the magnetic flux density, Wb/m}^2
\]
\[
H \text{ is the magnetic field, A/m}
\]
\[
\mu \text{ is the permeability of medium, H/m}
\]
\[
\mu_0 \text{ is the permeability of vacuum} = 4\pi \times 10^{-7} \text{ H/m}
\]
\[
\mu_r \text{ is the permeability of vacuum}
\]

**Magnetic moment** (Magnetic Resonance) The torque exerted on a magnet placed in a magnetic field of unit strength is called magnetic moment. Correspondingly the magnetic moment associated to a flat loop carrying a current and with its plane oriented parallel to the direction of a magnetic field of unit flux density is equal to the torque exerted on it. Any charged particle moving in a closed path produces a magnetic field which can be described at large distance as due to a magnetic dipole located at the current loop. The magnetic moment is a measure of the net magnetic properties of an object or particle.

The torque \( \tau \) on a magnetic dipole of moment \( \mu \) in a magnetic induction \( B \) is
\[
\tau = \vec{\mu} \times \vec{B}
\]
This means that the magnetic induction \( B \) tries to align the dipole so that the moment \( \vec{m} \) lies parallel to the induction. If no frictional forces are operating the work done by the turning force will be conserved. This gives rise to the following expression for the potential energy \( U \) of the dipole moment \( \mu \) in the presence of a magnetic induction \( B \):
\[
U = -\vec{\mu} \cdot \vec{B}
\]
A fundamental property of atomic nuclei is that those with odd atomic weights and/or odd atomic numbers, for example a nucleus of the hydrogen atom, which has one proton, possess an angular momentum or spin. Even if the nuclear spin is a property characterised only by quantum mechanics in the classical vector model the spin is visualised as a physical rotation similar to the rotation of a top about its axis. Nuclear magnetism of a nuclear spin system originates from the microscopic magnetic field associated with a
Magnetic polarity

A classical argument to justify the existence of this magnetic field is that a nucleus has electrical charges and it rotates around its own axis as if it has a nonzero spin (Figure M.4).

Physically it is represented by a vector quantity called nuclear magnetic dipole moment or magnetic moment. One fundamental relationship of particle physics is that the spin angular momentum and magnetic moment vectors are related to each other by

\[ \vec{\mu} = \gamma \vec{J} \]

where \( \gamma \) is a physical constant known as the gyromagnetic ratio and whose value depends on the nucleus, in particular on the ratio of electric charge over mass \( (e/m) \).

One of the consequences of having the magnetic moment proportional to the angular momentum is that an atomic magnet placed in a magnetic field will precess.

Related Articles: Magnet, Magnetic dipole

Magnetic resonance angiography (MRA)

The term angiography relates to imaging of vascular structures. Traditional angiography is performed with x-ray, using a contrast agent. MR angiography or MRA hence refers to MR examinations that aim to visualise blood vessels and related pathologies.

There are three main MRA techniques: time-of-flight MRA (TOF MRA), phase contrast MRA (PC MRA), and contrast enhanced MRA (CE-MRA), respectively. Common to all approaches is the use of 3D imaging sequences providing data over the vascular tree of interest. 3D data also enable the application of volume-rendering techniques such as maximum intensity projection (MIP), largely simplifying interpretation of the results.

TOF MRA relies on the fact that blood flowing into a slice has a higher signal than stationary tissue. Phase contrast MRA uses gradients to encode velocity information in the phase content of the image. These two techniques do not use any contrast agent. In CE-MRA, the signal of the blood is further enhanced via a magnetic contrast agent. See related articles for in-depth information on the different techniques.

Related Articles: Contrast enhanced MRA, Maximum intensity projection (MIP), MR angiography MRA, Phase contrast, Time of flight (TOF)

Magnetic resonance imaging (MRI)

MRI is the most commonly used abbreviation for the magnetic resonance imaging technique. Other abbreviations are, as an example, MRT (magnetic resonance tomography) or simply MR (magnetic resonance). MRI provides high quality images of the human body with high spatial resolution without using ionising radiation or invasive techniques. The inherent soft tissue contrast of the method provides differentiation in signal intensity between normal and diseased tissue. Recent developments of MR techniques encompass techniques for diffusion-weighted imaging (DWI), diffusion tensor imaging (DTI), perfusion-weighted imaging (PWI), functional MRI (fMRI) and MR spectroscopy (MRS) (Figures M.5 and M.6).

The impact of MRI on healthcare and diagnostic imaging has been tremendous, and the impact of MRI can be compared with the introduction of x-rays. The growth of MRI is also reflected in the number of installed units at hospitals throughout the world. The numbers continue to increase, and several countries in the world presently have more than 1 system per 100,000 inhabitants.

In 2003, Paul Lauterbur and Sir Peter Mansfield were awarded the Nobel prize in physiology or medicine 'for their discoveries concerning magnetic resonance imaging'.

The paths of development from the use of the NMR technique (see Nuclear magnetic resonance (NMR)) for the study of biological systems towards today’s MRI technique are indeed thrilling.

Attempts to measure – or at least investigate the possibilities for measurements of – NMR signals from living objects were made very early. Bloch inserted his finger into his Stanford spectrometer and obtained a strong signal while Purcell and Ramsey in

\[ + \]

FIGURE M.4 Spinning positively charged nucleus produces a magnetic moment \( \mu \).

FIGURE M.5 Magnetic resonance angiogram of the head, neck and upper thorax.

FIGURE M.6 Sagittal \( T_1 \)-weighted magnetic resonance image of the head.
1948 inserted their heads into the Harvard cyclotron without any observed biological contraindications on themselves.

At the same year as Hahn discovered spin echoes (leading to Carr and Purcell’s description of the 90°–180°-pulse), the first controlled measurement of the water NMR signal from living systems was made by Shaw and Elsken using a 15 MHz NMR system. The Swedish researcher Eric Odeblad played an important role in early studies of complex biological systems by determining relaxation times in animal tissue. Others working in the same area were Bratton and colleagues, and in the 1960s several groups gathered information regarding water $T_1$ and $T_2$ in living systems. Among the conclusions were that the relaxation times were decreased in living systems compared to the free water situation, and early attempts to create discrimination between different tissue states using NMR were made by Bratton and colleagues in 1965 and by Odeblad in 1968.

In 1971, Raymond Damadian found that relaxation rates in malignant rat cells differed from those in normal cells. This finding was rapidly reproduced by others, and may be seen as a starting point for the biomedical use of NMR. After attending an experiment performed by Hollis, Saryan and Morris, Paul Lauterbur started to investigate the possibilities of creating spatial resolution in an NMR experiment, and he successfully created the first two-dimensional NMR image using continuous wave (CW) NMR in 1973. The basic concept for the realisation of this image was that magnetic field gradients could be used to determine spatial position, since the signal in a well-defined small frequency interval must come from protons having this frequency range, and hence from protons at a specific position along the gradient axis. By 1D FT, the signals from different positions can be separated. In order to obtain 2D resolution, Lauterbur used a backprojection technique similar to the one developed for computerised tomography (CT). In 1973, independent of Lauterbur’s experiment, Sir Peter Mansfield used pulsed NMR techniques to obtain spatial resolution in one dimension using one discrete Fourier transform (DFT) of the signal from camphor crystal layers.

After the publication of these initial findings, Lauterbur continued his developments and rapidly showed 2D images in living objects, while Mansfield’s group developed a line-scan technique for 2D imaging, including a method for slice selection using a combination of RF excitation and a gradient in one direction and Waldo Hinshaw and colleagues, inspired by Lauterbur’s work, developed a ‘sensitive point’ method, in which a point in space was selected using switched gradients. All the early methods thus used the idea of spatial separation by magnetic field gradient application, but the goal – 2D resolved images – was reached in different ways. In 1975, Kumar, Welti and Ernst proposed a 2D/3D FT method for 2D/3D image reconstruction, and thereby created a base for the presently used so-called spin warp technique, which was described by Edelstein et al. in 1980. In the original proposal, durations of the encoding gradients were changed between different views, but in the spin warp technique gradient amplitude was changed instead. As an important part of the current technique, Hoult in 1977 suggested a gradient refocusing method in order not to lose signal during the slice selective process, and the same year Mansfield proposed the use of spin echoes in pulsed imaging experiments, as well as a technique for rapid sampling of raw signal data, the echo planar technique.

Following the first image of a living object by Lauterbur in 1974, images of larger objects became possible by building NMR units adapted to the new focus of visualising living animals and, eventually, humans. Hence, the first image of a human body part was published by Mansfield in 1977 using the line scan technique, and the first ‘whole-body’ NMR unit, developed by Raymond Damadian, an image of the human thorax was obtained the same year.

**Technique:** Medical MRI most frequently relies on the relaxation properties of excited hydrogen nuclei in water and lipase. When the object to be imaged is placed in a powerful, uniform magnetic field the spins of the atomic nuclei with non-integer spin numbers in the tissue all align either parallel or antiparallel to the magnetic field. In order to selectively image different voxels (volume picture elements) of the subject, orthogonal magnetic gradients are applied. Although it is relatively common to apply gradients in the principal axes of a patient (so that the patient is imaged in x, y and z from head to toe), MRI allows completely flexible orientations for images. All spatial encoding is obtained by applying magnetic field gradients which encode position within the phase of the signal. In one dimension, a linear phase with respect to position can be obtained by collecting data in the presence of a magnetic field gradient. In three dimensions (3D), a plane can be defined by ‘slice selection’, in which an RF pulse of defined bandwidth is applied in the presence of a magnetic field gradient in order to reduce spatial encoding to two dimensions (2D). Spatial encoding can then be applied in 2D after slice selection, or in 3D without slice selection. Spatially encoded phases are recorded in a 2D or 3D matrix; these data represent the spatial frequencies of the image object. Images can be created from the matrix using DFT. Typical medical resolution is about 1 mm$^3$, while research models can exceed 1 μm$^3$.

**The k-Space Formalism:** Since $x$ and $k$ are conjugate variables (with respect to the Fourier transform) we can use the Nyquist theorem to show that the step in $k$-space determines the field of view of the image (maximum frequency that is correctly sampled) and the maximum value of $k$ sampled determines the resolution, that is

$$\text{FOV} \approx \frac{1}{\Delta k}$$

Resolution $\approx |k_{max}|$

(These relationships apply to each axis ($x$, $y$ and $z$) independently.)

**MRI vs. CT:** A computed tomography (CT) scanner uses $x$-rays, a type of ionising radiation, to acquire its images, making it a good tool for examining dense tissue such as bone. MRI, on the other hand, uses non-ionising radio frequency signals to acquire its images and is best suited for non-calciﬁed tissue.

Both CT and MRI scanners can generate multiple two-dimensional cross sections (slices) of tissue and three-dimensional reconstructions. Unlike CT, which uses only $x$-ray attenuation to generate image contrast, MRI has a long list of properties that may be used to generate image contrast. By variation of scanning parameters, tissue contrast can be altered and enhanced in various ways to detect different features.

MRI can generate cross-sectional images in any plane (including oblique planes). CT is limited to acquiring images in the axial (or near axial) plane. However, the development of multi-detector CT scanners with near-isotropic resolution produces data that can be retrospectively reconstructed in any plane with minimal loss of image quality.

Magnetic resonance spectroscopy (MRS)

(Magnetic Resonance) This term refers to the use of magnetic resonance techniques to investigate chemical composition, as opposed to forming an image.

When a nucleus is placed in a static magnetic field in order to carry out a nuclear magnetic resonance (NMR) experiment, its resonance frequency depends primarily on the identity of the nucleus, via the equation $\omega = \gamma B_0$, where $B_0$ is the static field strength and $\gamma$ is the gyromagnetic ratio of the specific nuclear species. However, nuclei are surrounded by electrons, which partially shield them from the applied static field. Thus the static field experienced by a nucleus will differ from $B_0$, so that nuclei in all groups can generally not visible. However, lipids are frequently also present in tissues in a concentration that is orders of magnitude higher than that of most other hydrogen-containing compounds. Thus proton NMR signals from the body originate overwhelmingly from water, and signals from other compounds are generally visible in vivo by nuclear magnetic resonance. It is important to note that the peaks in an NMR spectrum arise from nuclei in a specific chemical group, but not from a specific compound. Thus a given peak may contain contributions from nuclei in the same group in a number of different compounds (e.g. the peak due to methyl protons in choline-containing compounds in proton MRS), while a single compound may give rise to multiple peaks if the molecule contains the same nucleus in more than one chemical group (e.g. the three peaks due to ATP in phosphorus MRS).

The potential clinical utility of MRS arises from the fact that the relative concentrations of different chemical compounds change in pathological states. For example, reduction in the level of N-acetyl aspartate in proton spectra from the brain is often indicative of neuronal loss, while reduction in citrate levels in spectra from the prostate is a marker for cancer. Using phosphorus MRS it is possible to probe cellular energetics and measure intracellular pH noninvasively. MRS has been variously dubbed the ‘noninvasive biopsy’, the ‘window on metabolism’ and as opening up ‘the new neurochemistry’. However, despite over 20 years of development few centres perform MRS examinations clinically on a routine basis as yet.

MRS has historical and methodological priority over MRI, which may be regarded as a form of proton MRS in which frequency differences arise as a result of the imposed gradients (and hence are related to spatial position) rather than from chemical shift. Chemical shift is a confounding effect in MRI, but its impact is minimised because water is present in tissues in a concentration that is orders of magnitude higher than that of most other hydrogen-containing compounds. Thus proton NMR signals from the body originate overwhelmingly from water, and signals from other compounds are generally not visible. However, lipids are frequently also present in high concentration, resulting in so-called chemical shift artefacts.

While MRI is invariably carried out using the hydrogen nucleus (proton), and specifically hydrogen in water, more generally MRS may be carried out using any nucleus that is present in sufficiently high concentration in the body, has an NMR-sensitive isotope with sufficient abundance, and has a sufficiently high sensitivity (dependent on the gyromagnetic ratio). In practice this limits in vivo natural abundance studies to hydrogen and phosphorus, but
carbon and fluorine MRS are feasible using exogenous compounds labelled with $^{13}$C and $^{19}$F respectively. When performing proton MRS in vivo, it is necessary to use special techniques to suppress the overwhelming signal from water and allow other peaks to be observed. This requirement meant that phosphorus techniques were exploited first, but proton MRS is now dominant because of the ease of implementation on a system designed for (proton) MRI.

In order to carry out MRS in vivo it is necessary to have some means of localising signal acquisition to the desired anatomical region. The ability to do this was a major breakthrough, since it allowed acquisition of spectra from specific tissues without excision from the body. Localisation may be achieved by using single voxel selection to focus on a specific volume of tissue, or by means of chemical shift imaging.

The resulting MRS spectrum can be quite complex, and sophisticated techniques are needed to identify the component peaks correctly and to analyse the spectrum, particularly if quantitative analysis is required.

**Related Articles:** Nuclear magnetic resonance (NMR), Chemical shift, Chemical shift artefact, Water suppression, Single voxel spectroscopy, Chemical shift imaging (CSI), Peak assignment, Spectral analysis


### Magnetic resonance tomography (MRT)

**Related Articles:** Magnetic resonance imaging (MRI), Magnetic resonance (MR), MRI

**See Nuclear magnetic resonance**

### Magnetic susceptibility

**Related Articles:** Inversion recovery, Inversion time (TI), Pulse sequence


### Magnetisation transfer contrast (MTC)

**Related Articles:** Inversion recovery, Inversion time (TI), Pulse sequence

**Related Article:** Spin echo

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**Magnetisation preparation**

*Magnetic Resonance* The term magnetisation preparation is frequently used to identify pulse sequence parts (see Pulse sequence) where the magnetisation is manipulated (prepared) prior to the readout of the signal in order to improve sequence performance with respect to signal-to-noise, contrast or acquisition time. A common type of magnetisation preparation is the application of a 180° pulse inverting the longitudinal magnetisation before applying a spin-echo, a fast spin echo or a gradient-echo sequence (see the following examples). Another type of preparation is applied on already existing transverse magnetisation, which is driven back to the longitudinal axis using a 90° pulse, enabling reduced repetition times in so-called driven equilibrium sequences.

#### EXAMPLE 1

By starting a pulse sequence with a 180° inversion pulse, thereby inverting the direction of $M_z$, and introducing a waiting time denoted TI before readout, objects with different $T_1$ will undergo different amounts of longitudinal relaxation during TI. In combination with a 90°–180° pulse train (a spin echo sequence), an inversion recovery experiment is created (see Inversion recovery) with an increased $T_1$ contrast range relative to a conventional 90°–180° spin echo sequence.

#### EXAMPLE 2

A 180° pulse can also be used in a fast gradient echo sequence. In this case, a fast train of low-flip angle RF pulses are applied after the inversion time, creating $T_1$ contrast in the resulting image. Using short repetition times and low flip angles, each $k$-space line is sampled with reasonably constant signal along the longitudinal relaxation curve (Figure M.8). This method, frequently denoted magnetisation-prepared rapid gradient-echo imaging (MP RAGE), enables fast three-dimensional imaging with $T_1$ contrast (1).
Magnetophosphenes

As myelinated white matter are strongly affected by MT and the MR signal in these tissues is decreased.

MT is often used to reduce the background in time-of-flight angiography and can also be used for better visualising multiple sclerosis (MS) lesions in T2-weighted and in gadolinium contrast enhanced T1-weighted images.

Magnetisation transfer effects may also introduce errors in certain applications. For example, in arterial spin labelling (ASL) the RF-pulse for inverting the spins before they enter the area of interest can also excite spins in the interesting area (bound protons). The exchange of protons between the bound and free pool will decrease the signal in the tissue that is not supposed to be affected, creating a false decrease in perfusion values.

Related Articles: Absorption, RF-pulse, T2


Magnetophosphenes

(*Magnetic Resonance*) Phosphenes are the flickering visual sensations caused by non-photic stimulation such as pressure on the eyes and mechanical shocks. They are generated in the retina and not in the optic nerve or the visual cortex. Gradient magnetic fields induced electrical currents which may excite the optic nerve and/or retina of the patient undergoing the MR examination at the threshold of perception. This excitation results in the visual sensation of flashes of light named magnetophosphenes (magnetic field-induced phosphenes). The same effect could be obtained also by rapid movements of the head or eyes in a 4 T static magnetic field while for lower strength of the magnetic field there is no induction of magnetophosphenes by head or eye movements. Magnetophosphenes have been elicited by current densities of as low as 17 μA/cm². It has been suggested that the threshold for the induction of magnetophosphenes is age dependant resulting in a lower value in younger subject. Since magnetophosphenes show the lowest threshold response to slow time-varying magnetic fields, it has been the practice to use their appearance in the production of guideline to limit exposure to MRI. The magnetophosphenes have never been regarded as harmful for the patients.


Magnetron

(*Radiotherapy*) Magnetrons are used in linear accelerators to produce high power microwaves, which are used to accelerate the
Magnification

(Diagnostic Radiology) The magnification of an image is a function of both the focal-to-film distance (FFD) and object-to-film distance (OFD); it is defined as

\[ m = \frac{FFD}{FFD - OFD} \]  

(M.2)

The image magnification is 1 when the object being imaged is placed on the film cassette surface, that is the OFD = 0.

As the OFD is increased the magnification increases, however, this also increases geometric unsharpness illustrated by an increased penumbra in Figure M.11. To overcome this, the focal spot is reduced as the magnification is increased. For example, in mammography the FFD is typically 60 cm, if the breast is offset by 20 cm the magnification will be 60/40 = 1.5. To reduce the increased geometric unsharpness produced by the magnification the focal spot size is reduced which reduces the penumbra.

**Abbreviations:** FFD = Focal-film distance, FOD = Focus-object distance and OFD = Object-film distance.

**Related Articles:** Geometric unsharpness, Focal spot, Focal-film distance Focus-object distance, Object-film distance, Penumbra

Main supply circuit, power supply

(General) A source of power suitable for operating of electrical and electronic devices, electrical power supply in most cases.

Alternating-current power supply is obtained from mains. Direct-current power supply is obtained from batteries, but can also be obtained by rectifying and filtering from alternating-current supply or by a power converter.

A typical design DC power supply that uses mains consists of a transformer, rectifier and a filter. Many supplies also have a voltage regulator at the output.

**Related Articles:** Line voltage, Voltage regulation

Mains frequency

(General) See Line voltage
Mammography (screen film)
(Diagnostic Radiology)

**Introduction:** Mammography is radiography of the breasts. It is significantly different from general radiography of other anatomical sites because of characteristics of the breast anatomy and pathologic conditions (Figure M.12).

**Equipment Design:** The breast is composed of soft tissue. Generally there is a background of adipose (fat) tissue which contains slightly higher density normal glandular structures and pathologic masses if they are present. The very small differences in density among the tissues result in low physical contrast and require an imaging procedure with high contrast sensitivity.

Small calcifications are important signs of some cancers, especially in the early stages when they are the most treatable. Visualisation of the small calcifications requires an imaging procedure with low blurring. Therefore, two of the characteristics that make a mammography system different from conventional radiography systems are the ability to form images with high contrast sensitivity and low blurring (Figure M.13).

The most common mammography system uses a molybdenum x-ray tube anode and a molybdenum filter. The x-ray spectrum produced with this combination gives a high contrast sensitivity that is generally optimised with radiation dose to the breast.

Some systems are designed with an alternative rhodium filter to be used with the molybdenum anode or an alternative rhodium anode track and rhodium filter combination. These are used to increase breast penetration, especially when imaging the more dense breasts.

The breast is compressed to produce a more uniform thickness, a thinner breast, and to reduce motion during the imaging exposure.

**Technical Factors:**

- **Exposure to the receptor** is generally controlled automatically for mammography. The necessary characteristics are a relatively wide exposure dynamic range (film latitude), high contrast transfer and low blurring (Figure M.14).
- **Diagnostic Radiology**

**Related Articles:**
- Operational amplifier, Voltmeter
- Film/screen
- Grid
- Receptor
- Mammography (screen film)

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**Mains voltage**
(General) See Mains supply circuit; Power supply

**Mains voltage drop**
(Diagnostic Radiology) See Voltage drop

**mA-metre**
(General) A mA-metre (milliampere metre) is an instrument used to measure electric current in an electric or electronic circuit. Electric current is measured in amperes (A). mA-metres are often found in x-ray equipment to measure the current in the anode circuit of the x-ray tube.

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Both film/screen and digital receptor/display systems are used for mammography. The necessary characteristics are a relatively wide exposure dynamic range (film latitude), high contrast transfer and low blurring (Figure M.14).

**Technical Factors:** The x-ray beam spectrum is controlled by the combination of anode material, filter, kV and mA. Selections are made based on breast size and density to optimise contrast sensitivity with respect to radiation dose. The selections are made either manually by the operator or automatically by some systems.

Exposure to the receptor is generally controlled automatically but can be adjusted by the operator for specific breast conditions. The larger of the two focal spots is used for general mammography and the small spot is used for the magnification technique which produces the least amount of blurring and the best visibility of detail, especially the small calcifications.

**High Contrast:** Two achieve high contrast the imaging is done at low kVp. This is usually in the range of 24–32kVp. This means that most of the contrast is generated using the photoelectric effect. To make the spectrum as narrow as possible a molybdenum anode with a molybdenum k-edge filter is the most common system. Some units have an additional rhodium filter and some also have an additional anode of rhodium or tungsten. This type of x-ray tube has low output so a 65 cm SSD is usually used. In addition a film with a high gamma is used to improve the contrast. This film is also low noise so small differences can be visualised.

**High Resolution:** In order to achieve high resolution a small focal spot is used. This is about 0.3 mm. In addition, to avoid cross-over effect, a single emulsion screen film system is used. The film is mounted in a cassette with the screen behind the film. The overall resolution is about 14lp/mm with the screen film system having a resolution of 20lp/mm.

**Mammography Unit:** The unit has a short SID because of the low output. The x-ray tube is tilted so to improve resolution. The tube is mounted so that the central ray passes through the chest wall edge of the breast instead of the centre of the receptor. This improves visualisation of all of the breast tissue. The unit has a grid of about 5/1. Mammography is done using many projections so the c-arm of the gantry rotates. The unit has a phototimer which is mounted behind the cassette. This is necessary because the photodetector cannot be made sufficiently radiolucent for mounting in front of the receptor. The unit is equipped with a breast compression system.
Man sievert
(Radiation Protection; General) This is the radiation unit given to collective effective dose. For more information, see the article on Collective dose.

Related Article: Collective dose

Manchester system
(Radiotherapy, Brachytherapy)

Manchester System – Intracavitary Brachytherapy for Cervix Cancer: The (intracavitary) Manchester system is based on the use of a combination of appropriately sized oovids and intrauterine tubes loaded in a standard way (originally Ra-sources). All combinations of oovids and intrauterine tubes are designed to give the same dose rate to the unique dosimetry point A, provided that the applicators are positioned correctly in the uterus and upper vagina. The Manchester system is a ‘time system based on the use of standard applicators’.

Two points, A and B, are defined in the classical Manchester system. Point A, the specification point of the Manchester system, is defined as a point 2-cm lateral to the uterine canal and 2-cm up from the mucus membrane of the lateral fornix of the vagina in the plane of the uterus. In clinical practice, point A is often defined 2-cm up along the axis of the central tube, from the lower end, and 2-cm away laterally (two points, left and right). Several definitions have been used in different centres, and the definition has in principle changed from an anatomy related to an applicator related point. The different definitions of the specification point make it difficult to compare results from different centres.

Point B is a reference point defined on the transverse axis through points A, 5 cm from the mid-line (two points, left and right). Point B is representative for the dose in the vicinity of the pelvic wall and for the dose to the obturator lymph nodes.

Manchester System for Interstitial Brachytherapy: The (interstitial) Manchester system, the Paterson–Parker system, aims at giving a ‘uniform dose’, that is a dose variation (outside the high-dose regions around each source) of about ±10%, to the target volume using radium needles. Sources of varying strength are used, the source strength distribution is non-uniform with more source strength at the periphery of the target volume and sources should be crossed at the ends. Tables are available giving the mg·h needed to deliver the specified doses for different implant sizes, provided that the implant rules are followed.

Related Article: Dosimetry system


Manganese
(General)

Symbol Mn
Element category Transition metal
Mass number A of stable isotope 30 (100%)
Atomic number Z 25
Atomic weight 54.9380 kg/kg-atom
Electronic configuration 1s² 2s² 2p⁶ 3s² 3p⁶ 4s² 3d⁵
Melting point 1519 K
Boiling point 2334 K
Density near room temperature 7210 kg/m³ (7.21 g/cm³)
**History:** Manganese is found freely in nature. In prehistoric times the element was adopted as a pigment in paint and it was also later used by Egyptians and Romans in glass-making. Nowadays manganese has many industrial uses, most notably as a component in a range of alloys (including stainless steel) and as a cathode material in disposable batteries.

**Medical Applications:** Defibrillation: Manganese dioxide button cells have a relatively stable voltage during discharge, such that they are employed in the medical field for defibrillation of the heart.

**Manual afterloading**  
*(Radiotherapy, Brachytherapy)*

**Source Handling and Loading:** The brachytherapy source/s must be handled and loaded into the applicators for treatment, and many methods have been used over the time. These methods have been developed primarily to reduce the dose to the personnel, but also to improve the quality of the treatment itself.

Manual afterloading is used when

1. Applicators, needles, catheters, etc., are inserted
2. Correct applicator positions are verified using dummy sources
3. It is necessary to improve the accuracy in applicator positioning, as there is no risk of dose to staff
4. The sources are inserted into the applicators manually

See *Source loading in brachytherapy*

**Related Articles:** Brachytherapy, Source loading in brachytherapy, Manual loading, Manual afterloading, Remote afterloading, Remote afterloading unit

**Manual loading**  
*(Radiotherapy, Brachytherapy)*

**Source Handling and Loading:** The brachytherapy source/s must be handled and loaded into the applicators for treatment, and many methods have been used over the time. These methods have been developed primarily to reduce the dose to the personnel, but also improving the quality of the treatment itself.

Manual loading is used in historical methods, for example when handling of radium sources, which were manually introduced and removed; treatment times vary from a number of hours to several days. It was also used when speedy insertion techniques were mandatory and when the patient was a radiation source during the long treatment.  

See *Source loading in brachytherapy*

**Related Articles:** Brachytherapy, Source loading in brachytherapy, Afterloading, Manual afterloading, Remote afterloading, Remote afterloading unit

**mAs selector**  
*(Diagnostic Radiology)* The mAs selector differs from the mA selector by the fact that it is also linked to the Timer of the HVG of an x-ray equipment. This way, the mAs selector selects and measures the quantity of charge \( Q = mA \times s \) during the exposure, which directly corresponds to the dose during the exposure. For this purpose the anode current is integrated over the time of exposure (using an integrator). The eventual timing error in the mAs devices should be related to the fact that 25% change in mAs of an exposure is clearly visible (and often related to exposure point in some of the exposure tables used in the radiography practice).

**Related Articles:** High voltage generator, mA selector, Filament circuit, Exposure point

**Mask mode fluoroscopy**  
*(Diagnostic Radiology)* Mask mode fluoroscopy is one of the initial names of a method used digital subtraction angiography (DSA). In this most common method subtraction is performed between the two images – the second one (with contrast) and the first one (without contrast), called mask.

The other DSA method is Time interval difference subtraction, where the mask is not fixed, but changes in time (i.e. subtraction is performed between the image with contrast and one of the previous images with less contrast). This method gives information about the changes of contrast in time.

The third main DSA method is K-edge subtraction, where the two images are made with different kV – one just below the K-edge of the contrast media, the other – just above this K-edge. In fact this is dual energy subtraction, and as the contrast medium is most often Iodine, the two energies are around its K-edge.

**Related Article:** Digital subtraction angiography

**Mass attenuation coefficient**  
*(Radiation Protection)* The amount of attenuation in an absorber is related to the mass or density of the absorbing material. In terms of its mass, the attenuation coefficient is given by

\[
\mu / \rho
\]

where

- \( \mu \) is the linear attenuation coefficient
- \( \rho \) is the density of the material

**Related Articles:** Attenuation, Linear attenuation coefficient

**Mass collision stopping power**  
*(Radiotherapy)* The mass collision stopping power is the rate of energy lost by the charged particle resulting from the sum of the soft collisions, causing the excitation and ejection of an electron which carries a relatively small energy, and the hard collisions that result in the ejection of an electron with relatively large energy transfer (delta ray).

For electrons the mass collision stopping power is given by

\[
\left( \frac{dE}{\rho dx} \right)_{\text{coll}} = \frac{N_{\text{A}}Z_{\text{eff}} r_{0}^{2} 2m_{e} c^{2}}{A \beta^{2}} \left( \ln \left( \frac{E_{k}}{I} \right) + \ln(1 + \tau / 2) + F(\tau) - \delta \right)
\]

where

- \( N_{\text{A}} \) is the Avogadro’s number
- \( Z \) is the atomic number of the medium
- \( r_{0} \) is the classical electron radius
- \( m_{e} \) is the electron rest mass
- \( c \) is the speed of light in vacuum
- \( A \) is the atomic mass of the medium
- \( \beta = v/c \) is the speed of the particle relative to \( c \)
- \( E_{k} \) is the kinetic energy
- \( I \) is the mean excitation energy of the medium
- \( \tau = E_{k}/m_{e} c^{2} \)
- \( \delta \) is the density effect correction

**Mass defect**  
*(Nuclear Medicine)* Mass defect refers to the apparent loss of mass when particles form atoms. The mass defect stems from the fact that a bound system has a lower energy than the unbound constituents...
and therefore has a lower mass than the sum of its particles in a free state. When free particles are binding, the excessive energy is released as thermal vibrations or photon emissions, that is the mass is ‘transported’ to another location.

**Mass energy absorption coefficient**

*(Radiation Protection)* The energy absorption coefficient, $\mu_e$, is a linear function of the density of the medium. This dependence can be avoided using the mass energy absorption coefficient, $\mu_{m,e}$, defined as follows:

$$\mu_{m,e} = \frac{\mu_e}{\rho}$$

where

$\rho$ is the density of the material

$\mu_{m,e}$ has dimension $L^2/M^1$ and is usually given in cm$^2$/g

The radiation intensity attenuation factor associated with absorption, $\exp[-(\mu_{m,e} \cdot x)]$, where $t$ is the absorber thickness, can be rewritten as

$$\exp[-(\mu_{m,e} \cdot x)]$$

where $x$, defined as the product $x = \rho \cdot t$, is the so-called mass thickness.

The mass energy absorption coefficient can be used to calculate the absorbed dose in a medium. For instance, if the radiation intensity in a point of the medium, $I [\text{W/(cm}$cm$^2$ s)$]$, and the energy absorption coefficient, $\mu_{m,e} [\text{cm}^2$/g$]$, are known, the product $(\mu_{m,e} \cdot I)$ gives an estimate of absorbed dose at that point.

The mass energy absorption coefficient does not eliminate the dependence on radiation energy and on the composition of the medium; hence, in practical conditions an average mass energy absorption coefficient is needed.

**Related Articles:** Energy absorption coefficient, Absorbed dose, Average mass energy absorption coefficient

**Mass number**

*(Nuclear Medicine)* The mass number is the sum of protons and neutrons in an atomic nucleus. All atoms of a certain element have the same number of protons but the number of neutrons can differ, that is same element but different mass numbers. Atoms of the same element but with different mass numbers are referred to as isotopes.

**Mass of radium**

*(Radiotherapy, Brachytherapy)* Calibration of source strength is a very important part of a comprehensive brachytherapy quality system. The instruments, ion-chambers and electrometers, used for source strength determinations should have calibrations that are traceable to national and international standards.

Specification of source strength for photon emitting sources:

Source strength for a photon emitting source can be given as a quantity describing the radioactivity contained in the source or as a quantity describing the output of the source:

1. Specification of contained activity
   a. Mass of radium in mg
   b. Contained activity in Ci or GBq

2. Specification of output
   a. Equivalent mass of radium; mg Ra eq
   b. Apparent activity
   c. Reference exposure rate
   d. Reference air kerma rate
   e. Air kerma strength

When brachytherapy was introduced as a treatment modality, the only sources available were radium sources. Source strength was given as the mass of the radium contained in the encapsulated source. The filtration of the encapsulation was also given; usually 0.5 mm Pt for needles and 1–2 mm Pt for tubes. (The UK’s National Physical Laboratory [NPL], a standards/measurement institute established at the turn of the last century, acquired its first radium standard in 1913, made by Marie Curie, and specified in terms of mass of Radium.)

In modern brachytherapy dosimetry, reference air kerma rate or air kerma strength is the quantity used to calculate absorbed dose. (Kerma is the kinetic energy released in matter.)

See Source strength for a full description of specification of source strength.

**Related Articles:** Source strength, Contained activity, Equivalent mass of radium, Apparent activity, Reference air kerma rate (RAKR), Air kerma strength

**Mass radiative stopping power**

*(Radiotherapy)* The mass radiative stopping power is the rate of energy lost by the charged particle resulting in the production of bremsstrahlung radiation.

For electrons the mass radiative stopping power is given by

$$\left( - \frac{dE}{d\rho dx} \right)_{\text{rad}} = \sigma_0 N A Z^2 c^2 E_k A E_k = \frac{E_k}{Z}$$

where

$N_A$ is the Avogadro’s number

$Z$ is the atomic number of the medium

$m_e$ is the electron rest mass

$c$ is the speed of light in vacuum

$A$ is the atomic mass of the medium

$E_k$ is the kinetic energy

$\sigma_0$ in the range of energies 0.5–100 MeV varies between 5.33 and 15:

$$\sigma_0 = \alpha \left( \frac{E_k}{4\pi m_e c^2} \right)^2 = 5.80 \cdot 10^{-8} \cdot \text{cm}^2/\text{atom}$$

where $\alpha$ is the fine structure constant.

From this relationship it can be seen that the energy loss by bremsstrahlung increases directly with $Z$, the atomic number of the medium and more slightly with the electron energy.

**Mass stopping power**

*(Radiotherapy)* The average linear rate of energy loss per unit of path length $x$ is called stopping power. Dividing the stopping power by the density of the medium it results in a new dosimetric quantity called mass stopping power. Common units for the mass stopping power are MeV cm$^2$/g. In the mass stopping power the dependence on the density medium is removed except for the polarisation effect. The mass stopping power does not differ greatly for materials with similar atomic composition. In addition the mass stopping power in a gas is independent on the pressure because dividing the stopping power by the density exactly compensates for the pressure. The mass stopping power could be subdivided into a mass collision stopping power and a mass radiative stopping power:

$$\left( - \frac{1}{\rho} \frac{dE}{dx} \right)_{\text{rad}} = \left( - \frac{1}{\rho} \frac{dE}{dx} \right)_{\text{coll}} + \left( - \frac{1}{\rho} \frac{dE}{dx} \right)_{\text{rad}}$$

**Related Articles:** Mass collision stopping power, Mass radiative stopping power
**Matching layer**

(Ultrasound) The matching layer improves the sensitivity in ultrasonic transducers by making the sound transmission more efficient. This is achieved by reducing the reflections in the boundary between transducer and tissue.

The matching layer has an acoustic impedance value between the impedance of the transducer element and the soft tissue and should ideally be: $Z_{m} = \sqrt{Z_{t} \cdot Z_{s}}$ where $Z_{m}$ is the acoustic impedance for the matching layer, $Z_{t}$ is the impedance of the transducer and $Z_{s}$ is the impedance of the soft tissue.

If the thickness of the matching layer is exactly a quarter of a wavelength then 100% of transmission through the transducer/tissue boundary is possible. What happens in the transducer is that the ultrasound beam reverberates inside the matching layer and the reflected beams coming back into the transducer disc will cancel out and the beams coming into the tissue are reinforced. This is due to that a reflection from a medium with lower acoustic impedance will have a 180° phase shift of the pulse, but a reflection from a medium with higher impedance will have no phase shift (Figure M.15).

Many transducers have multiple matching layers to provide efficient transmission for a spectrum of ultrasound frequencies. With a well-designed matching layer the demands of the backing layer will be reduced.

**Related Articles:** Backing layer, Reflection coefficient

**Matrix array**

(Diagnostic Radiology) Matrix arrays are used in diagnostic imaging technology when a large number of either detectors or display elements are needed to be addressed simultaneously. In diagnostic radiology matrix arrays are used in both flat panel displays and x-ray detectors. Matrix arrays are classed as either active or passive.

Passive matrices are only used in old monochrome liquid crystal display (LCD) devices such as older laptop and mobile phone displays. They are small matrix devices of limited contrast and refresh rate. The matrix is formed by a grid of electrodes with each pixel element formed by the overlapping regions of the electrodes (Figure M.16a). Row electrodes are referred to as scanning electrodes and column electrodes as data electrodes. To address a specific pixel the row electrode is switched to ground (0 V) and the column electrode is set to the desired voltage, $V$ (Figure M.16b). The display is addressed row-by-row. Passive arrays suffer several major setbacks: (1) Although the selected pixel is set to the correct voltage all other pixels in the column are also given a small voltage which partially switches them on, this is referred to as cross-talking. In the example of a $3 \times 3$ matrix shown in Figure M.16b the selected pixel is given a voltage $V$, whereas the surrounding pixels acquire a voltage $V/2$. (2) Once a pixel row has been addressed, the state is not maintained while other rows within the matrix are addressed and the pixel state will degrade as the root mean squared of the pixel capacitance. (3) The refresh rate of the display is slow. Due to these limitations passive matrices are limited in size and have a low contrast ratio. They are not used for medical imaging; instead, active matrices are used.

As opposed to passive matrices, each element within an active matrix contains a switching element (usually a thin film transistor, TFT). This switching element allows every element to be individually addressed and for the element state to be maintained when other elements are being addressed. Active matrices can be produced as a large area matrix (currently in excess of $40 \times 40$ cm$^2$), which allows it to be used as a fundamental constituent in modern digital x-ray detectors and modern LCD displays. In medical imaging technology when a large number of either detectors or display elements are needed to be addressed simultaneously. In diagnostic radiology matrix arrays are used in both flat panel displays and x-ray detectors. Matrix arrays are classed as either active or passive.

**Related Articles:** Backing layer, Reflection coefficient

**FIGURE M.15** Use of matching layer (M) between two media (PZT and soft tissue) with differing acoustic impedances. The reflected waves are cancelled by destructive interference between wave 3, 4 and 6. (Graphs courtesy of EMIT project, www.emerald2.eu)

**FIGURE M.16** (a) Passive matrix of $3 \times 3$ elements. (b) Circuit diagram for the passive matrix, the selected pixel 11 is the given voltage $V$ while the non-selected pixels experience cross-talk and are given a voltage of $V/2$. (a) Column electrodes
imaging detectors the active matrix array is used for both direct and indirect radiography. Arrays used for both types of imaging incorporate a two-dimensional array of imaging pixels, which consists of a switching element used for data read-out (typically a TFT) and a sensing and storage element.

The active matrix utilises thin film technology which allows the deposition of hydrogenated amorphous silicon (a-Si:H), making it ideal for construction of both TFTs and photodiodes. Large area arrays are formed by plasma deposition of thin layers of the appropriate materials (e.g. amorphous silicon) onto a glass substrate, once deposited they can then be etched to the desired pattern by a process called photolithography (Figure M.17).

Figure M.17 shows a typical array used in a medical imaging flat panel detector. Within the array each pixel consists of the switching element and an element to detect incoming photons and store them as charge. The image read-out process is controlled by altering the voltage applied across the switching element. Firstly, to allow each pixel to detect a signal during exposure the voltage across each switching element is set to an ionisation or ‘off’ state. The signal is then read-out by changing the switching voltage row-by-row to the conducting or ‘on’ state which allows the charge stored in each pixel to be drained by the charge collector electrode and passed to the multiplexer. The voltage change is controlled by the gate line driver. As the read-out process is controlled by the external circuitry, each row of pixels requires a separate control line driver to alter the switching voltage, and each column its own amplifier. This process is called the active matrix read-out.

The active matrix array allows the radiographic image signal to be read-out sequentially, line by line. Fluoroscopic images are acquired in real-time by permitting all other rows that are not being read-out to continue to detect the incoming signal during exposure.

**Abbreviations:** AMA = Active matrix array and TFT = Thin film transistor.

**Related Articles:** Thin film technology (TFT), Amorphous silicon, Flat panel detector, Liquid crystal display (LCD), Active matrix array, Active matrix liquid crystal flat panel display


**Matrix array**

(Ultrasound) The term matrix array describes an array in which the elements are arranged both along the main access of the transducer and also in the elevation plane so that better control of the transmitted and/or received ultrasound beam in the elevation plane can be achieved. Matrix arrays can be used to improve slice thickness in a conventional B-mode image or to provide volume acquisition for 3/4 D displays and multiplanar displays.

Matrix arrays are sometimes categorised as 1.25-, 1.5- and 2D arrays. In 1.25D arrays, outer elements are switched on to improve focusing in the far field and switched off for the near field (Figures M.18 and M.19). If focusing is only required in receive mode, then dynamic aperture control can be performed electronically by incorporating outer elements in the received signal from deeper tissue.

**Figure M.17** Active matrix array and peripheral electronics.

**Figure M.18** 1.25D, 1.5D and 2D matrix arrays have different functions. In 1.25D and 1.5D arrays, the additional rows of elements are used to improve control of slice thickness in a conventional array.

**Figure M.19** In conventional arrays (left) focusing in the slice thickness is achieved by using an acoustic lens. With a 1.25D array, each element is subdivided. Inner elements are used for focusing at near depth (middle) and the outer elements are used to improve focusing at depth (right).
Matrix size

*Diagnostic Radiology* The matrix size of a digital image is the number of pixels contained in the image. This is often expressed by giving the number of pixels in each dimension or direction of the image. Matrix size has a direct effect on two important image characteristics; pixel size and the numerical size or number of bytes to record the image.

Pixel size is a factor in determining the spatial resolution or detail of the imaging system. When an image is digitised into a matrix of pixels each pixel becomes an additional source of blurring that adds to the other sources within the system, such as focal-spot size and blurring within the receptor.

Pixel size is the ratio of the image size, or field of view (FOV), to the matrix size. Increasing the matrix size decreases the pixel size and therefore improves the resolution and visibility of detail. Respectively the visualisation of a larger FOV with the same matrix leads to decreased resolution.

Usually the matrix of medical imaging systems is square – that is with equal number of pixels in X and Y direction. A digital radiographic system with a matrix size of 2048 × 2048 pixels can produce a pixel size of approximately 0.2 mm (400/2048) for its maximal image field size (400 × 400 mm). This provides a spatial resolution of 2.5 lp/mm. If the image field of view is smaller (e.g. 200 × 200 mm), the pixel size will be decreased and the spatial resolution will be increased. If the system matrix size is doubled to 4096 × 4096, the pixel size will be 0.1 mm (400/4096) and the spatial resolution would be 5 lp/mm.

If a CT scanner has a 512 mm scanning diameter and the matrix is 1024 × 1024 pixels, the pixel size will be 0.5 mm (512/1024) and the spatial resolution will be 1 lp/mm. However if the scanner has collected sufficient number of projections, then the raw data from this scan can be used for subsequent reconstruction of another smaller image. For example, a region of interest (ROI) with diameter 128 mm can be reconstructed. In this case the pixel size of the final image will be 0.125 mm (128/1024), and a spatial resolution of 4 lp/mm.

The previous example assumes a new reconstruction of the scan from the collected projections. If ROI is applied without this (i.e. display zoom only), the spatial resolution will be the same, but the pixels will be optically enlarged on the monitor.

Another dimension or parameter of a digital image in addition to matrix size (the number of pixels) is the depth of each pixel expressed as the number of bits to record each pixel. This parameter relates to the contrast characteristics of the image. If the pixel has 8 bits it will be able to present 2^8 = 256 levels of grey (or colours).

There is a difference between the bit depth (bits per pixel) needed and used to acquire images and for the display of images. A relatively large pixel bit depth (16 bits per pixel) might be used to record both digital radiographs and CT images. This gives a large exposure dynamic range. The images that are then processed and windowed for display do not require as many bits per pixel.

Contemporary digital images used in medicine have 16 bits of depth, where usually 12 bits are used to record the image contrast, and the other 4 bits are used for supporting information (e.g. text or graphs displayed over the image). These 12 bits present 2^12 = 4096 levels of grey (or colours) and are more than enough for the human visual system. 4096 levels of grey is also completely sufficient for various densitometric measurements (measurement the optical density of the pixel, corresponding to the radiation absorption of the respective voxel from the anatomical object).

A final example – an image with a numerical size of 2048 × 2048 × 16 (also called 4 mega pixel matrix) will contain approximately 67 Mega bits. If the bit depth is reported in bytes (1 byte = 8 bits), the raw image file size will be 8 MB.

Maximum dose

*Nuclear Medicine* The maximum dose refers to the location with highest registered absorbed dose. The term is typically used in
dosimetric calculations, for example internal and external radiotherapy. In non-quantitative measurements the dose is usually normalised to the location with highest dose.

**Maximum frequency follower**

*(Ultrasound)* The method for obtaining the maximum and minimum frequency followers (frequency envelopes) from the sonogram has been described by Gibbons et al. (1981) and improved by Evans et al. (1989a,b) and consists of comparing the spectrum of the Doppler shift signal with a threshold and marking the first and the last points where the spectrum is greater than the threshold. The main advantage of using the maximum frequency follower rather than the mean frequency follower is that the former is less affected by the thump wall filter, which removes power at low frequencies both from wall movements (which is the reason why it is used) and from slow moving blood (an undesirable but unavoidable artefact). Furthermore, a comparison of the behaviour of the mean and maximum frequency allows a qualitative estimation of the flow profile: with parabolic flow the mean frequency should be close to half the maximum while for plug flow both followers will be approximately the same (Evans et al. 1989a,b).

Since the Doppler shift produced by a moving target is proportional to the velocity of the target, and since there are a large number of targets (scatterers) moving with a range of velocities in a blood vessel, the Doppler shift signal is a wide-band signal containing information about the velocity distribution of blood within the sample volume.

The best way to obtain information from the Doppler shift signal is to perform real-time spectrum analysis, enabling the assessment of the evolution in time of the blood velocity distribution and the usual way of displaying the resulting information is the Doppler sonogram (link).

From each of the individual spectra that make up the sonogram one can obtain the mean frequency (proportional to the mean velocity), which is the ‘normalised first moment’

\[ \bar{f} = \frac{\sum_{i=0}^{N-1} P_i \times i}{\sum_{i=0}^{N-1} P_i} \]  

(M.3)

where

- \( P_i \) is the intensity of each individual frequency bin
- \( i \) is the counter of frequency bins in the Fourier transform of the Doppler shift signal \( \Delta f(t) \)

Joining the points obtained from Equation M.3 over time produces the intensity weighted mean frequency follower and, provided that the angle of insonation is known, the mean velocity follower.

The sonogram on Figure M.22 is showing maximum velocity envelope and mean velocity envelope, as well as fiducial markings for start of systole, peak velocity and end of diastole (cross markers).

**Further Readings:** Doppler ultrasound, Sonogram

**Related Articles:** OSEM, Reconstruction

**Maximum likelihood expectation maximum (MLEM)**

*(Nuclear Medicine)* The most commonly used iterative reconstruction methods in commercially available systems are the maximum-likelihood expectation maximum method (commonly denoted the MLEM) and its accelerated version called ordered-subsets expectation-maximum (OSEM). In practice, the equation that describes the process is given by

\[ I_{new}^{new} = \frac{\sum_{j}^{h_{ji}} P_j}{\sum_{j}^{h_{ji}} h_{ji}} \sum_{k}^{P_{kj}} h_{kj} I_{old}^{old} \]

where

- \( I \) is the image to be created
- \( P \) is the measured projection
- \( h_{ji} \) is the probability (sometimes called the transfer matrix) that the pixel
- \( i \) will contribute to the projection bin \( j \)

In its simplest form that assumes no photon attenuation, no scatter contribution and collimator blur, \( h_{ji} \) takes the value of unity along the ray-of-view for the current projection angle. This then reduces the formula to

\[ I_{new}^{new} = \sum_{j}^{h_{ji}} P_j \sum_{k}^{P_{kj}} h_{kj} I_{old}^{old} \]

The summation term under \( I_{new}^{new} \) is needed because the backprojection step is a summation step and therefore a normalisation with the number of projection angles is essential to keep the number of counts in the reconstructed images the same as have been acquired.

**Related Articles:** OSEM, Reconstruction

**Maximum (minimum) intensity projection**

*(Diagnostic Radiology)* Maximum intensity projection (MIP) and minimum intensity projection (mIP) are volume rendering techniques for image display. A two- or three-dimensional model of CT data set (or part of it) can be created by displaying the maximum or minimum CT numbers encountered along the viewing angle. These techniques are used, for example, for CT angiography (MIP) or...
Maximum (minimum) intensity projection (MIP) for optimal visualisation of contrast differences between different tissues.  

**Related Article:** Volume rendering  

### Maximum (minimum) intensity projection (MIP)  
*(Magnetic Resonance)* The MIP is a very useful image rendering technique for three-dimensional MR angiographic data. A projection line is defined through the slab at a certain angle, and the projection value is given by the maximum pixel intensity along the projection line. In this way, bright structures form the projection image, suppressing background with lower signal intensity. This property makes the MIP well suited for both TOF MRA and contrast-enhanced MRA (Figures M.23 and M.24), as these techniques visualise blood vessels as bright.  

Clinically, a set of radial projection lines are often defined, and scrolling through the resulting images gives the impression of viewing the vascular tree from different angles.  

An analogue to the maximum intensity projection is the minimum intensity projection, often abbreviated MinIP.  

**Related Article:** Magnetic resonance angiography (MRA)  

### Maximum permissible concentrations  
*(Radiation Protection)* Maximum permissible concentration is the maximum quantity per unit volume of a radioactive material in air, water and foodstuffs that is not considered an undue risk to human health.  

In order to evaluate the maximum permissible concentration in case of occupational exposure to radionuclides, ingestion and inhalation dose coefficients shall be taken into account. This means the committed effective dose per unit intake via ingestion corresponding to different gut transfer factors (i.e. the proportion of the intake transferred to the body fluid in the gut) for various chemical forms; and the committed effective dose per unit intake via inhalation for the default lung absorption types (fast, moderate and slow). The aforementioned quantities are calculated taking into account the latest models for the respiratory tract, the biokinetic models for systemic activities and the appropriate transfer factors for the components of the intake cleared from the lungs to the gastrointestinal tract.  

The inhalation and ingestion dose coefficients and all the necessary parameters for the calculation of the occupational exposure in relation to all known radioisotopes in use are provided by International Commission on Radiological Protection (ICRP) and International Atomic Energy Agency (IAEA) publications. The various concentrations present will determine the number of working hours in the given conditions.  

In case of public exposure, the same methodology is applied, taking into account the different dose limits for the public and the total permanence in the environment. Different age groups, namely less than 1, 1–2, 2–7, 7–12, 12–17 and more than 17 years, are considered. Increased transfer factors are used for infants from 0 to 3 months.  

The maximum permissible concentration allowed for each radionuclide depends on the earlier-explained parameters and effects on the human body. The total amount of radionuclides in the working place and in the environment are established taking into account the fact that the dose limits for occupationally exposed workers and for public, respectively, shall be respected. As a result the competent authorities (at local and national levels) are responsible for issuing authorisations, monitoring the working conditions and checking the environment in order that the maximum permissible concentrations of radioisotopes in food, air and water are compatible with the maximum permissible amounts in the body.  


### Maximum permissible dose (MPD)  
*(Radiation Protection)* Up to the 1950s, all dose limits set for workers occupationally exposed to ionising radiation was based on avoiding deterministic effects. In other words, workers were not allowed to reach a threshold dose. No consideration was taken in
setting dose limits for the risk of causing cancer or other stochastic effects in the population of exposed workers. From 1934, the limit set was called the maximum permissible dose.

Evidence emerging from the survivors of the atomic bombs at Hiroshima and Nagasaki, together with the follow-up of patients undergoing radiotherapy for non-malignant disease such as ankylosing spondylitis, and subsequently developing cancer, convinced the international community that a dose limit merely set to avoid deterministic effects was not good enough.

In 1956, ICRU proposed a revised complete system of radiation quantities to describe radiation exposure and resultant dose to a human, together with a set of radiation units. These were exposure, unit the Roentgen, radiation absorbed dose to measure the deposition of energy in tissue, unit the rad, and the Roentgen equivalent man to measure the equivalent whole body effect of a partial or series of partial body exposures, unit the rem.

The revision of the system of units by ICRU then prompted ICRP a year later to issue new advice on dose limitation, revising the concept of the maximum permissible dose (MPD) to take full account of both the avoidance of deterministic effects, and a consideration of the limit of acceptability of exposing the working population to a stochastic risk of cancer.

In defining this new limiting dose, they also determined that the annual MPD should be reduced to 5 rem (equivalent to 50 mSv in the modern unit, that is 2.5 times the current dose limit for radiation workers). They also introduced the concept of cumulative dose into the dose limits by specifying the annual MPD in terms of age at exposure with the formula

\[ \text{MPD} = 5(N - 18) \text{ rem} \]

where \( N \) is the age of the worker at exposure.

This formula indicated that radiation workers could not be less than 18 years of age. Above that age their annual dose had to be an average of 5 rem over their working lifetime.

**Related Articles:** Roentgen, Absorbed dose, Deterministic effects, Stochastic effects

### Maximum target absorbed dose

*(Radiotherapy)* The International Commission on Radiation Units and Measurements (ICRU) recommends a common method of dose specification which could be generally adopted to permit a comparison between the treatment practices. Normally a non-uniform dose distribution is obtained in the target volume and therefore for practical reasons it is useful to report specific doses such as the maximum target absorbed dose.

The maximum target absorbed dose is the highest absorbed dose in the target area that can be regarded as ‘clinically meaningful’. The latter term implies that at least a minimum area is irradiated to the dose level designated as ‘maximum’. The minimum area recommended for this purpose is 2 cm², unless the whole target area is less than 4 cm², in which case a minimum area of 1 cm² should be taken to define the maximum target absorbed dose.

The ICRU indicates the value 2 cm² considering this area as the smallest one for which the absorbed dose can be calculated with confidence and also because the maximum absorbed target dose is often related to the limiting effects of treatment.

**Related Articles:** Mean target absorbed dose, Minimum target absorbed dose, Modal target absorbed dose, Median target absorbed dose, Hot spots

**Further Reading:** ICRU (International Commission on Radiation Units and Measurements). 1978. Dose specification for reporting external beam therapy with photons and electrons. ICRU Report 29, Washington, DC.

### Maximum velocity

*(Ultrasound)* See Maximum frequency follower

### Maxwell gradients

*(Magnetic Resonance)* In an idealised application of a gradient field in MRI, a linear magnetic field strength variation is introduced along one direction in the bore of the scanner. However, Maxwell’s equations demonstrate that it is not possible to produce an isolated linear variation in just one direction. Additional gradients in the orthogonal directions called ‘Maxwell’ or ‘concomitant’ gradients will also exist.

With the application of a gradient of amplitude \( G \) in the \( z \)-direction, the \( z \) component of the field \( H \) at point a distance \( z \) from the isocentre is given by

1. \( H_z = H_0 + Gz \)

where

2. \( H = (H_x, H_y, H_z) \)

The divergence of the vector \( H \) is

3. \( \nabla \cdot \vec{H} = \frac{\partial H_x}{\partial x} + \frac{\partial H_y}{\partial y} + \frac{\partial H_z}{\partial z} \)

From Maxwell’s equations,

4. \( \nabla \cdot \vec{H} = 0 \)

so

5. \( \frac{\partial H_x}{\partial x} + \frac{\partial H_y}{\partial y} = G \)

The rate of change of field with position (i.e. the gradient) along a plane orthogonal to the applied gradient is seen to be non-zero. These gradients are the Maxwell or concomitant gradients. Maxwell gradients are a source of small but unwanted spatial variations in the resonance frequency of spins in an imaged volume.


### Mayneord \( F \) factor

*(Radiotherapy)* Mayneord \( F \) factor is an equation derived based on inverse square law to correct for the change in percentage depth dose values as a result of changes in focus to skin (FSD) distance in radiotherapy treatment set-up.

For instance, the per cent depth dose of a radiotherapy beam at depth \( d \) in a patient is \( %DD_1 \), if the focus to skin distance of the set-up changes from FSD1 to FSD2, the per cent depth dose value \( %DD_2 \) of the new set-up can be determined using the following equation:

\[
F = \left( \frac{\text{FSD}_2 + d_m}{\text{FSD}_1 + d_m} \right)^2 \times \left( \frac{\text{FSD}_1 + d}{\text{FSD}_2 + d} \right)^2
\]

where \( F \) is the Mayneord \( F \) factor.

The equation only takes into account the changes due to inverse square law effect. It does not account for any changes in scattering dose. For this reason, \( F \) factor works well for small fields. The factor
has been used in manual treatment planning for accommodating FSD changes of up to about 40% with acceptable error.

**Maze**

*Radiation Protection* Radiotherapy suites are designed to protect staff and the general public from primary and scattered radiation. The entrance to the room needs special consideration, and will either use a heavily shielded mechanical door, or a maze. Figure M.25 shows a typical radiotherapy room layout including an entrance maze. A correctly designed maze will prevent primary and reduce scattered radiation reaching the room entrance, including neutrons. In general, if the maze offers at least two reflections, then this will provide sufficient protection for 10MV photons. In most cases a maze eliminates the need for more shielding at the entrance, which removes the need for a large heavy door. Some kind of interlocked barrier is still recommended to prevent free access to the room, for example a waist high gate. This design of room is considered more patient friendly and enables faster access to the patient and room in an emergency. However the maze must be big enough for beds and equipment to be physically wheeled round.

The provision of a maze may be difficult where an existing room is upgraded to a higher energy unit, and a protective door may be the solution. Alternatively, if there is a pre-existing maze that needs to be improved for a higher energy unit, then there are a number of measures that can be incorporated:

- Extra baffles and lintels can be used to effectively increase the length of the maze.
- A 5% Boron or Lithium loaded polyethene lining can be used to reduce the energy of fast neutrons.
- A light interlocked door can be placed at the inner maze entrance, but used only for the higher energy treatments.

**MDR (medium dose rate)**

*Radiation Protection, Brachytherapy* See Medium dose rate (MDR)

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**Mean absorbed dose to air**

*Radiation Protection* This is purely a qualitative term used to define the mean absorbed dose to air in conditions when the radiation field is variable.

**Related Article:** Absorbed dose

**Mean dose per cumulated activity**

*Nuclear Medicine* The mean dose per cumulated activity $S$ is used to calculate the radiation dose received by a target organ $r_i$ from activity located within a source organ $r_j$. The procedure is often complicated and time consuming, especially when there is a large number of source organs and multiple emission types to consider. The quantity $S$ has the unit ‘Gy/Bq s’. $S$ is calculated for a number of source–target pairs and for several radionuclides. $S$ is determined by the amount of energy emitted by the cumulated activity in the source organ ($\Delta$) and the anatomic relationship between the two organs and the target organ composition ($\Phi$). The mean dose per unit cumulated activity is

$$S(r_i \leftarrow r_j) = \frac{1}{N} \sum_{i} \Phi_i (r_i \leftarrow r_j) \Delta_i \quad \text{(M.4)}$$

Given $S$ and the cumulated activity $\tilde{A}$ the total radiation dose $D$ to a target organ from a specific source organ is

$$D(r_i \leftarrow r_j) = \tilde{A} \times S(r_i \leftarrow r_j) \quad \text{(M.5)}$$

**Related Articles:** Cumulated activity, Equilibrium absorbed dose constant, Absorbed fraction, MIRD formalism


**Mean electron energy**

*Radiation Protection* The energy spectrum of an electron beam depends on the spread in energy of the intrinsic linac beam and the modification by the energy loss and scatter in any material through which the beam passes. In Figure M.26, the typical distribution of the electron beam energy is shown before leaving the accelerating structure of the linac [graph (a)]; at the phantom surface [graph (b)]; and at a depth $\tau$ in the phantom [graph (c)]. Due to the complexity of the spectrum there is no single parameter which can characterise the electron beam; therefore several parameters are defined such as the maximum energy $E_{\text{max}}$, the modal energy $E_p$ and the mean energy $\bar{E}$.

The mean electron energy as a function of depth in the phantom is needed in some dosimetric protocols for the choice of correction parameters for absorbed dose measurements with an ion chamber measurement. The Harder’s relation is employed to estimate the mean electron energy:

$$\bar{E}_d = \bar{E}_0 \left(1 - \frac{d}{R_p}\right)$$

where

- $E_0$ is the average incident electrons energy at the phantom surface
- $d$ is the depth of measurement
- $R_p$ is the practical range in the same units
The stochastic quantity is defined as the quotient of $dE/dm$, energy imparted per unit volume of absorber.

Mean energy imparted

(Radiation Protection) Absorbed dose is a macroscopic quantity defined as the quotient of $dE/dm$, energy imparted per unit volume of absorber.

At a microscopic level energy absorption is a stochastic process. The stochastic quantity is $z$, the specific energy imparted, where

$$z = \frac{\varepsilon}{m}$$

where $\varepsilon$ is the energy imparted in an absorber of mass $m$, in a specific event. The formal definition of absorbed dose is

$$D = \frac{d\varepsilon}{dm}$$

where $d\varepsilon$ is the mean energy imparted in joules (J).


Mean free path

(Radiation Protection) As a beam of photons passes through an absorber, it is attenuated, due to collisions with the molecules within the material. The number of photons in the beam decreases according to the relationship

$$n(x) = n(0)e^{-x/l}$$

(M.6)

where

- $l$ is the mean free path
- $n(0)$ is the number of photons per second in the beam passing $x = 0$
- $n(x)$ is the number of photons that have travelled an additional distance, $x$, without being scattered out of the beam

Mean free path may also be calculated using the following equation:

$$l = \frac{1}{n n r^2}$$

(M.7)

where

- $l$ is the mean free path
- $n$ is the number of molecules
- $r$ is the collision radius of the molecule

If the absorber is a gas at atmospheric pressure, the mean free path is equal to several hundred atomic diameters.

Related Articles: Absorber(s), Mean free path for photons in attenuation


Mean glandular dose

(Radiation Protection) The dose to which patients are exposed to during mammography procedures is normally defined in terms of mean glandular dose (MGD) measured in mGy, rather than in terms of effective dose. Breast tissue is relatively highly radiosensitive, and patients may be healthy and subject to exposure as part of a screening program. Therefore, doses during mammography must be optimised so that they are as low as reasonably practicable to minimise the risk of cancer in the exposed population, whilst ensuring that the resultant images allow early diagnosis of possible breast cancers such that overall prognosis for those patients is greatly improved.

Calculation of MGD is described in IPEM Report 89 and uses a standard breast model based on the work of Hammerstein et al. (1979) and Dance et al. (1990, 2000a). A 4.5-cm thick Perspex phantom of semi-circular cross-section in the horizontal plane of diameter 16 cm is used, and the incident air kerma measured.

The measured air-kerma is related to the MGD for a standard breast using conversion factors, applied as shown in Equation M.7. The entrance air kerma for a 4.5-cm thick Perspex phantom is equivalent to that for a 5.3-cm thick breast with a glandularity of 29% in the central region. The model also has 0.5 cm thick adipose layers at the top and bottom and has been found to be typical for breasts of this compressed thickness for patients aged 50–64 years:

$$D = K_{45} \cdot g_{53} \cdot c_{53} \cdot s$$

(M.8)

where

- $D$ is the mean glandular dose
- $K_{45}$ is the entrance kerma for 4.5-cm thickness of Perspex
- $g_{53}$ is the g-factor for the 5.3-cm thick standard breast
- $c_{53}$ is the conversion factor to allow for the glandularity of the 5.3-cm thick standard breast
- $s$ is the spectral correction factor

The g and c factors are dependant upon the half value layer of the spectra used and can be estimated using the tables shown in IPEM Report 89.
**Mean life**

*(Nuclear Medicine) See Average lifetime of atoms*

**Mean target absorbed dose**

*(Radiotherapy)* The International Commission on Radiation Units and Measurements (ICRU) recommends a common method of dose specification, which could be generally adopted to permit a comparison between the treatment practices. Normally a non-uniform dose distribution is obtained in the target volume and therefore for practical reasons it is useful to report specific doses such as the mean target absorbed dose.

For its calculation, it is necessary to calculate the dose in a large number of discrete points (lattice points), uniformly distributed in the target area. The mean target absorbed dose is then calculated as the mean of the absorbed dose values at these points. Mathematically,

$$\frac{1}{N} \sum_{i,j} D_{i,j}$$

where

$N$ is the number of the points in the matrix

$D_{i,j}$ is the dose at lattice point $i, j$ located inside the target area ($A_T$)

**Related Articles:** Maximum target absorbed dose, Minimum target absorbed dose, Median target absorbed dose, Modal target absorbed dose, Hot spot

**Further Reading:** ICRU (International Commission on Radiation Units and Measurements). 1978. Dose specification for reporting external beam therapy with photons and electrons. ICRU Report 29, Washington, DC.

**Mean transit time (MTT)**

*(Magnetic Resonance)* The mean transit time (MTT) is the mean time that it takes for a non-diffusible tracer (representing the blood) to pass through a microvascular or capillary system from the arterial to the venous side. In normal brain tissue MTT is on the order of 4–6 s. The function $h(t)$ describes the distribution of times required by different tracer molecules when passing through the system (following an instantaneous tracer input). MTT can be calculated using the central volume theorem: For a system with blood flow $F$, the blood volume $V$ is the sum of all paths taken by the tracer within the microvascular system. If the product $h(t)dt$ is the tracer fraction that leaves the system between time $t$ and $t + dt$, and this tracer fraction has passed through a volume $Ft$, the total blood volume $V$ is given by the summation of all $Ft$ volumes weighted by the tracer fraction $h(t)dt$ at time $t$:

$$V = \int_0^\infty (Ft)h(t)dt = F \int_0^\infty th(t)dt$$

The MTT is given by

$$\text{MTT} = \int_0^\infty th(t)dt = \frac{V}{F}$$

It may also be useful to remember that the tissue residue function $R(t)$, that is the fraction of tracer that remains in the tissue at a given time $t$ following an instantaneous bolus, is given by

$$R(t) = 1 - \int_0^t h(t)dt$$

which implies that $\frac{dR(t)}{dt} = -h(t)$. By using an integration by parts of the function $1 \cdot R(t)$ the following relationship is obtained:

$$\int_0^\infty R(t)dt = \lim_{t \to \infty} [tR(t)] + \int_0^\infty t \cdot h(t)dt$$

Since the volume must be finite, it is evident that $[tR(t)] \to 0$ when $t \to \infty$ [1]. Hence, MTT can be calculated as

$$\text{MTT} = \int_0^\infty R(t)dt$$

**Further Reading:** Cerebral blood flow, Cerebral blood volume, Perfusion imaging

**Related Articles:** Extrapolation ionisation chamber, Ionisation chamber

### Measuring chamber

*(Radiation Protection)* The measuring chamber is an air-filled ionisation chamber (Figure M.27) used for the measurement of gamma radiation exposure. The ionisation charge is proportional to the exposure and the ionisation current to the exposure rate.

It is based on the Bragg–Gray principle:

$$D_m = W \times S_m \times P \text{ (Gy)}$$

where

$D_m$ is absorbed dose in a given material

$W$ is equal to the average energy loss per ion pair in the gas

$S_m$ is a relative mass stopping power of the material to that of the gas

$P$ is a number of ion pairs formed in the gas per unit mass

A gas-filled ionisation chamber can be used for the indirect measurement of the absorbed dose in an arbitrary medium.

**Related Articles:** Extrapolation ionisation chamber, Ionisation chamber

![Figure M.27](image) Scheme of an air-filled ionisation chamber.
Mechanical index (MI)

(Ultrasound) The mechanical index is an indicator of the relative risk of mechanical bioeffects resulting from diagnostic ultrasound.

Bioeffects from ultrasound can be broadly divided between thermal and mechanical effects. Possible mechanical effects of ultrasound include streaming of fluid and cavitation. Cavitation is known to occur at outputs much higher than diagnostic ultrasound, for example in lithotripsy. Although there is no evidence of adverse mechanical effects occurring in clinical practice using diagnostic ultrasound, there is concern that there should be limits on ultrasound output to ensure safe practice.

The output display standards (ODS) were put forward by the American Institute of Ultrasound in Medicine (AIUM) and the National Electrical Manufacturers Association (NEMA) in 1992 as a guide for users to monitor the output level, and by association, relative risk of ultrasound scanners.

The risk of adverse mechanical bioeffects increases with increased rarefractional pressure (also known as negative pressure) and decreases as ultrasound frequency increases.

The mechanical index is a non-dimensional index defined as

$$MI = \frac{P_{2/3}(f_{\text{max}})^{1/2}}{C_{\text{MI}}}$$

where

- $P_{2/3}$ is the peak derated rarefraction pressure using 0.3 dB/cm MHz (derated for energy absorption by tissue)
- $f_{\text{max}}$ is the ultrasound working frequency
- $C_{\text{MI}} = 1 \text{ MPa MHz}^{-1/2}$ where $C_{\text{MI}}$ is a correction factor for the model tissue target being considered

The upper limits set by the Food and Drug Administration (FDA) in their track-3 guidance are

- MI = 0.23 for ophthalmic applications
- MI = 1.9 for all other applications

MI and TI (thermal index) are generally displayed on the ultrasound screen as an aid to the user. MI has been found to be useful in setting levels of output for contrast agent use. Low MI contrast agents can be used with low outputs to enable continuous scanning to highlight flow in small arteries. High MI devices can be used to disrupt the contrast agent bubbles enabling techniques to measure volume re-filling using interval scanning.

Related Articles: Isocentre, Radiation isocentre

Mechanical locking system

(General) A mechanical locking system is any system which limits or prevents access to some physical space such as a room or the inside of an equipment cabinet. Such a lock may also be fitted to an electrical power switch to prevent unauthorised use.

This may be anything from a simple pin driven through a cog locking it onto its shaft, to a complex electro-mechanical safety interlock system to prevent physical access to a radiotherapy treatment bunker during irradiation.

A practical lock needs at least one stable state – the locked position requiring significant mechanical force, tool, or code to change it to the unlocked position. Such a lock may be activated by chance or by default, and where this might pose a problem a ‘deadlock’ having two stable states is preferable.

Mechanical transducer

(Ultrasound) The term mechanical transducer is used to describe a transducer where movement of the transducer elements is used to form the image by ‘sweeping’ through a plane or volume. Most conventional commercial systems use array transducers where the elements are fixed and beams are steered electronically. Mechanical transducers are still used for the following:

- 2D B-mode imaging. The transducer is mechanically swept around an axis and a sector image provided from a single element (Figure M.28).
- 3D imaging. A curvilinear array is mechanically swept in the elevation plane providing a series of images from a swept volume.
- High frequency (>20 MHz) imaging. Difficulties in constructing small arrays have led researchers to use single element transducers moved in a line or plane to provide a plane or volume of ultrasound data.

**Further Readings:**

Mechanical interlock

(Radiation Protection) See Interlock

Mechanical isocentre

(Radiotherapy) The mechanical isocentre is the point about which the various moveable parts (gantry, collimator, treatment couch) of the treatment machine rotate. While the isocentre is ideally a point, in reality it is typically a small sphere of radius less than 1 mm within which the various axes of rotation intersect.

To verify the mechanical isocentre, a front pointer must first be inserted into the head of the linear accelerator. The collimator is then rotated and the walkout of the front pointer observed compared to a piece of graph paper attached to the treatment couch. Typically it should be possible to achieve a radius less than 0.5 mm. The walkout of the front pointer should also be checked as the couch is rotated, again typically on the order of 0.5 mm. Finally, the position of the tip of the front pointer with respect to a fixed point should then be checked at the four main orthogonal gantry angles, and the deviation should be less than 1 mm.

**Related Articles:** Isocentre, Radiation isocentre

![Diagram of a single element transducer](image-url)

**FIGURE M.28 Diagrammatic representation of a single element transducer which is rotated to provide a swept area which is used to produce a sector scan.**
Medical physics

(General: Clinical) Directional anatomical terms describe the relationship of structures relative to other structures or locations in the body. 'Medial' means towards the mid-line, middle, away from the side (example, the middle toe is located at the medial side of the foot).

See Anatomical relationships

Median target absorbed dose

(Radiotherapy) The International Commission on Radiation Units and Measurements (ICRU) recommends a common method of dose specification which could be generally adopted to permit a comparison between the treatment practices. Normally a non-uniform dose distribution is obtained in the target volume and therefore for practical reasons it is useful to report specific doses such as the median target absorbed dose.

The median target absorbed dose is the central value among the set values of the absorbed dose at all the lattice point in the target area when arranged according to magnitude.

Related Articles: Mean target absorbed dose, Maximum target absorbed dose, Minimum target absorbed dose, Modal target absorbed dose, Hot spots


Medical physics

(General) Medical physics is a profession which deals with the application of the concepts and methods of physics for the prevention, diagnosis and treatment of human diseases with a specific goal of improving human health and well-being. A specific definition of the International Organization for Medical Physics (IOMP) gives the following description for the professional activities of a medical physicist:

Medical physicists apply knowledge and methodology of science of physics to all aspects of medicine, to conduct research, develop or improve theories and address problems related to diagnosis, treatment and rehabilitation of human disease. They are directly involved with patients and people with disabilities. Tasks include (1) conducting research into human disorders, illnesses and disabilities and investigating biophysical techniques associated with any branch of medicine; (2) conducting specialised examinations of patients and the disabled, improving patient care and clinical services and developing innovative imaging and non-imaging diagnostic procedures for specific medical applications; (3) developing novel instrumentation and physiological measurement techniques, mathematical analysis and applications of computers in medicine in response to clinical need for patients and aiding to everyday living for the disabled; (4) ensuring the quality, safety testing and correct maintenance and operation of treatment machines, x-ray equipment, radiation treatment planning computers; medical uses of ultrasound, MRI and infrared and the correct delivery of prescribed radiation doses to patients in radiation therapy; (5) ensuring the accuracy of treatment unit parameters and settings used for a patient’s treatment, including correct transfer of parameters between the simulator, treatment plan and the treatment unit, and periodic review of each patient’s chart; (6) calculating dose distributions and machine settings, design and fabrication of treatment aids and treatment-beam modifiers for individual patient treatments; (7) in vivo measurement to verify the dose delivered to a patient; participation at patient discussion conferences; (8) advising and consulting with physicians on the physical and radiobiological aspects of patients’ treatments, and the development of treatment plans in such applications as use of ionising radiation in diagnosis, therapy, treatment planning with externally delivered radiation as well as use of internally implanted radioactive sources given the state of technology; (9) planning, directing, conducting and participating in supporting programs and remedial procedures to ensure effective and safe use of ionising and non-ionising radiation and radio nuclides in human beings by physician specialist; (10) formulating radiation protection guides and procedures specific to hospital environment and other professional groups and organisations and conducting specialised measurements and producing protocols to minimise radiation exposure of patients, staff and the general public; (11) participating in and contributing to the development and implementation of national and international standards, laws and regulations relating to patient safety, particularly to radiation and radioactive materials; (12) teaching principles of medical physics to physicians, residents, graduate students, medical students, technologists, and other health-care professionals by means of lecturers, problem solving and laboratory sessions; (13) preparing, publishing and presenting scientific papers and reports and (14) supervising and managing radiation workers and other health professional workers.


Medical physics expert (MPE)

(Radiation Protection) The medical physics expert is defined in the European Directives as: ‘an expert in radiation physics and technology whose training and competence to act is recognised by competent authorities and who, as appropriate, acts or give advise on patient dosimetry, on the development and use of complex techniques and equipment on optimisation, on quality assurance, including quality control, and on other matters relating to radiation protection within the scope of the Directive’.

The Directive further states, ‘In radiotherapeutic practices, a medical physics expert shall be closely involved. In standardized therapeutic nuclear medicine practices and in diagnostic nuclear medicine practices, a medical physics expert shall be available. For other radiological practices, a medical physics expert shall be involved, as appropriate, for consultation on optimization including patient dosimetry and quality assurance including quality control, and also to give advice on matters relating to radiation protection concerning medical exposure, as required’.

Individual national legislation and guidance implementing the Directive provides more detailed information on the education, training and experience required to be appointed as an MPE and the duties to be undertaken.

The fields of competence of an MPE are normally radiotherapy physics, diagnostic radiological physics, nuclear medicine physics and medical health physics; an MPE is competent and authorised to practice independently in one or more of these fields. In addition to these fields of competence in some cases the MPE is also involved in other clinical and biomedical activities.


Medium dose rate (MDR)

(Radiotherapy, Brachytherapy)

Dose Rates in Brachytherapy: Different dose rates are used in brachytherapy treatment techniques. The International Commission on Radiation Units and Measurements, ICRU, defined these dose

1. Low dose rate, LDR
   a. 0.4–2.0 Gy/h
   b. Traditional technique; 0.5 Gy/h, 60 Gy with treatment time 5 days
   c. Large amount of clinical data
   d. (NOTE: Ultra low dose rate 0.01–0.3 Gy/h)

2. Medium dose rate, MDR
   a. 2–12 Gy/h
   b. Seldom used

3. High dose rate, HDR
   a. > 12 Gy/h = 0.2 Gy/min
   b. Treatment times approx. 5–20 min (comparable to external beam therapy)
   c. Clinical data available

4. Pulsed dose rate, PDR
   a. Mimics LDR, using many small ‘HDR pulses’ during a longer treatment time example: 1 pulse per hour during 24 hours, 0.5 Gy per pulse given in 5 min; total dose 12 Gy per day

The radiobiological effects in the tissues irradiated depend on the type of applicator used, on the fractionation scheme and on both dose and dose rate distributions. As stated in the ICRU Report 38, ‘the clinical experience accumulated with radium techniques cannot be applied to new irradiation conditions without careful consideration’. This includes consideration of both tumour effects and effects on normal tissues.

**Abbreviation**: ICRU = International Commission on Radiation Units and Measurements.

**Related Articles**: Brachytherapy, Dose rates in brachytherapy, see also articles under radiobiology


**Medium frequency portable x-ray machine**
*(Diagnostic Radiology)* See High frequency generator

**Mega electron volt**
*(General)* See Electron volt

**Meisberger polynomial**
*(Radiotherapy, Brachytherapy)*

**Point Source Calculations**: Dose distributions around brachytherapy sources are dominated by the inverse square law behaviour. Different slowly varying functions have been used to describe the deviation of the dose distribution from $1/r^2$, and the Meisberger polynomial is one of them. This polynomial was published in 1968, and coefficients were given for the brachytherapy sources used at that time; gold, iridium, caesium, radium and cobalt.

The Meisberger polynomial has been used for a long time in treatment planning systems to characterise the radial behaviour of cylindrical brachytherapy sources.

**Related Articles**: Source models, Point source calculation


**Metal artefact**
*(Diagnostic Radiology)* Metal objects in the patient, such as dental fillings, prostheses, surgical clips and electrodes, can cause severe streaking artefacts in the image (especially in computed tomography). Sometimes these are called ‘star artefacts’ – Figure M.29. The artefact can be caused by a combination of low signal, beam hardening, partial volume, undersampling at sharp interfaces and limited dynamic range of the detectors and display.

Metal artefacts can be eliminated by removing of metal objects, where possible, or avoidance of them, for example, by gantry angulation. Otherwise metal artefact reduction software may be available to reduce their appearance. This software identifies projection data which has ‘over-ranged’ due to high attenuation of the metal object. This data is then replaced by interpolating data from both sides of the metal object (Figure M.30). Images showing the effectiveness of metal reduction software are shown in Figure M.31.

Some manufacturers use an extended CT number scale to reduce metal artefacts caused by over-ranging of attenuation values.

**Related Articles**: Artefact, Beam hardening, Cone beam artefact, Helical artefact, Image artefact, Motion artefact, Partial volume effect (artefact), Ring artefact

**Metal x-ray tube**
*(Diagnostic Radiology)* In order to diminish the metalisation of the glass, some manufacturers produce x-ray tube envelopes whose central part is made of metal (steel, which is insulated and grounded). The end parts of this envelope are still made of glass and vacuum inside is similar.

In more recent designs the entire envelope is made of metal (using black ceramic insulation inside and additional ceramic insulators at the high voltage connections). A typical ceramics is beryllium oxide due to its very high thermal conductivity. Often the metal x-ray tubes are also called ceramic x-ray tubes.

The metal tube envelopes have also better heat dissipation abilities and are more robust. They can be used with anode dics with mass on the order of 2000 g (i.e. with double heat storage ability) what make them especially useful for angiography and CT scanners.

The metal tubes use a composite (double) exit window – made of beryllium and aluminium (the x-rays first pass through Be, and after this through Al). These tubes have smaller inherent filtration as firstly Be absorbs less radiation than the glass and secondly they can
be placed closer to the housing, what diminishes the oil absorption. The inner side of the metal envelop is specially coated to absorb the infrared radiation from the heated anode. At present more and more x-ray tubes use this type of envelope (Figures M.32 and M.33).

**Related Articles:** Glass envelope, Filament heating, X-ray tube, Anode, Liquid metal bearing

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**Magnetic Resonance** Magnetic field-related translational attraction and torque present hazards to individuals with certain implants or devices. The presence of an intracranial aneurysm clip made from ferromagnetic materials is contraindicated for MR procedures because magnetically induced forces may displace these clips causing serious injury to the patient or his death. The risk of the clip displacement can be a minimal risk in other parts of the body because after some time fibrosis and encasement of the clip help to keep it in a stable position. The pulsed radio frequency (RF) magnetic fields,
which are used to obtain the MR signals from tissue induce electrical currents in conductive metal implants and determine their heating. The magnitude of the increased heating of tissues due to the presence of the metallic implant depended on the dimensions, orientation, shape and location of the metallic implant in the patient. This increased heating of surrounding tissues primarily concentrates in a small volume and can produce burst. Metallic implants produce also susceptibility artefacts in magnetic resonance imaging. The artefacts are related with the metal characteristics and the orientation of the implant in the magnetic field. The artefact can therefore be minimised by optimally positioning patient in the magnet.

Related Article: Implant

Metastable nucleus
(Nuclear Medicine) A metastable nucleus is an excited nucleus ‘trapped’ in an excited state. The metastable nucleus will eventually de-excite down to its ground state. In the periodic table nomenclature an m is placed after the atomic number to mark that the nucleus is metastable. A common metastable nucleus used in nuclear medicine is $^{99m}$Tc; the metastable technetium nucleus has a half-life of approximately 6h, meaning that half the excited states will have de-excited after 6h. A 140keV photon suitable for emission imaging is generated in the de-excitation process.

Metastable state
(Nuclear Medicine) See Metastable nucleus

Metastasis
(Nuclear Medicine) Metastasis refers to cancer cells that have detached themselves from the primary tumour and colonised in another organ or tissue. The new cancer colony is referred to as a metastasis.

A detached cancer cell (or cell cluster) can spread through the body via the blood stream and the lymphatic system. If cells from a primary lung cancer spread to and metastase in the liver, they are then categorised as metastatic lung cells rather than liver cancer.

Microbubbles
(Ultrasound) Contrast agents for ultrasound are made up of microbubbles that dramatically change the acoustic properties of the medium and thereby provide a stronger signal from blood. More details and detection strategies are provided under Contrast agents.

Microdosimetry
(Radiation Protection) The deleterious effects of ionising radiation occur as stochastic events at a sub-cellular level with dimensions on the order of microns. Conventional quantities for measuring radiation dose and radiation quality are absorbed dose (D) and linear energy transfer (LET). These quantities are measured over macroscopic distances.

Microdosimetry, as the name suggests, is concerned with the deposition of energy at microscopic levels. It is important in developing an understanding of how and why biological effects may differ following irradiation by the same absorbed dose and radiation quality. Microdosimetric events are stochastic in nature and there are two principal microdosimetric quantities, lineal energy (y) and specific energy (z).

Lineal energy is the quotient of the energy imparted, $\varepsilon$, in a single event and the mean chord length, $l$, of the volume in which the deposition occurred; thus,

$$y = \frac{\varepsilon}{l}$$

Lineal energy is measured in units of keV/μ and differs from LET, a non-stochastic quantity, in that it has no energy cut-off value.

Specific energy (z) is the quotient of energy imparted, $\varepsilon$, and the mass of the volume in which the event occurred; thus,

$$z = \frac{\varepsilon}{m}$$

The mean specific energy, $\bar{z}$, which is the expectation value of a specific energy between $z$ and $z + dz$ is a non-stochastic quantity and equivalent to the absorbed dose, D.

Lineal energy and specific energy differ in that the former is the deposition from a single event whereas the latter may result from more than one event.

Measurements of microdosimetric quantities are made with proportional chambers. These are gas-filled devices with components made from tissue equivalent materials. They are designed to be Bragg–Gray cavities imitating tissue dimensions on the order of microns.

Related Articles: Absorbed dose, Linear energy transfer

Micro-MLC
(Radiotherapy) For small fields such as those used for brain tumours or boost fields in the head and neck, better resolution of the field margins may be required than for larger PTVs. Several miniature multileaf collimators have been developed to be used for these cases with 1.5–6mm leaf widths projected at the linac isocentre.

Micro-MLCs (sometimes referred as mini MLCs) have been typically configured as self-contained accessories that can be attached to the collimator of a linear accelerator for specific treatment techniques and removed for conventional use of the machine. Fibre-optic transmission lines are used to communicate with a PC-based digital control system. The secondary jaws of the accelerator are set to a fixed field size during the use of the micro MLC so that the leaves of the micro MLC need be only long enough to cover a reduced maximum field size (Figure M.34).

Another non-conventional MLC system is the MIMiC device provided by NOMOS Corporation. It is designed to collimate the x-ray field to a fan-beam that is dynamically modulated by short-stroke leaves as the gantry of the accelerator is rotated. The modulated fan beam irradiates a transverse plane of the patient that is 2-cm thick. Each leaf pair has only two positions: completely open or completely closed. This system is sometimes referred to as bimodal MLC.

Abbreviation: MLC = Multileaf collimator.

Related Articles: Multileaf collimator, Stereotactic radiosurgery

FIGURE M.34 $m^{3}$ high-resolution multileaf collimator produced by BrainLab (www.brainlab.com/download/pdf/BrainLABIGTBrochureEnglish.pdf).
Microwaves

Microwaves are electromagnetic waves (i.e. energy waves propagated by paired, transversely oscillating electric and magnetic fields) and are part of the electromagnetic spectrum. The microwave part of the spectrum is characterised by wavelengths on the order of approximately 1 mm – 1 m, frequencies of between roughly 0.3 and 300GHz. Microwaves are in the non-ionising part of the electromagnetic spectrum.  

**Related Article:** Electromagnetic energy spectrum

Midpoint dose

(Radiotherapy) The midpoint dose is the dose delivered at the midline of a patient treated by two photon beams directed along the same axis from opposite sides of the treatment volume. If the parallel opposed beams are equally weighted and normalised to the midpoint value, the dose distribution can be made uniform within the irradiated volume depending on the patient thickness and the beam energy and flatness. The dose near the patient surface increases compared to the midpoint dose when the patient thickness increases or the beam energy decreases. In Figure M.35 the depth dose curves for parallel opposed beam normalised to the midpoint dose value are reported for different beam energies. The so-called tissue lateral effect consists in the increase of the surface dose relative to the midpoint dose.

The ratio between the maximum peripheral dose and the midpoint dose is reported in Figure M.36 for different patient thickness and beam energies.


**Related article:** Système Internationale (SI)


**Hyperlink:** http://en.wikipedia.org/wiki/Microprocessor

**Micropet**

(Nuclear Medicine) A pre-clinical detector system for imaging the radionuclide distribution in small animals. The spatial resolution in a micro-PET system is primarily determined by the positron physics, namely the positron path length. A conventional PET scanner has better resolution than a conventional SPECT but for the micro systems the relationships is reversed. The obvious medical benefit with a small animal PET imaging system is that radiopharmaceuticals can be tested on animals and evaluated before clinical trials.

**Related Article:** Micro-SPECT

**Microprocessor**

(General) The microprocessor is now one of the main components of every computerised system. While it is usually a single chip central processing unit (CPU) – usually 16-bit, 32-bit or 64-bit design, special microprocessors are used for digital signal processors (DSP) and graphics processing units (GPU). These microprocessors are main components of any imaging system.

**Hyperlink:** http://en.wikipedia.org/wiki/Microprocessor

**Micro-schut**

(General) A micro-switch is an electro-mechanical switch that needs very little force for closing or separating the electrical contacts. They are often used in control and regulation circuits where a movement of the operator or a mechanical tool controls a process. It is often used as a stop switch in medical electro-mechanical devices and medical imaging equipment that has movable parts.

**Micron**

(General) Micron (symbol μ) is a unit of length corresponding to micrometre (symbol μm) in the International System of Units (SI):

\[ 1 \mu = 1 \mu m = 10^{-6} \text{ m} \]

**Related article:** Système Internationale (SI)


**Hyperlink:** http://en.wikipedia.org/wiki/Micron

**Micro-PET**

(Nuclear Medicine) A pre-clinical detector system for imaging the radionuclide distribution in small animals. The camera parameters are related in a similar way as for a large scale SPECT system. System resolution, for example is primarily determined by the bore diameter of the SPECT system which gives a superior resolution compared to a conventional SPECT system. There is also the obvious medical benefit of trying new radiopharmaceuticals on animals instead of patients. A micro-PET can be used together with a micro-PET, CT or MR scanner in a multimodality system.

**Related Article:** Micro-SPECT

**Micro-SPECT**

(Nuclear Medicine) A pre-clinical detector system for imaging the radionuclide distribution in small animals. The camera parameters are related in a similar way as for a large scale SPECT system. System resolution, for example is primarily determined by the bore diameter of the SPECT system which gives a superior resolution compared to a conventional SPECT system. There is also the obvious medical benefit of trying new radiopharmaceuticals on animals instead of patients. A micro-SPECT can be used together with a micro-PET, CT or MR scanner in a multimodality system.

**Related Article:** Micro-PET

Minification gain

*Diagnostic Radiology* See Total brightness gain

Minimum detectable activity (MDA)

*Nuclear Medicine* The minimum detectable activity (MDA) (also called minimum detectable amount of activity), of a radionuclide may be defined as the smallest activity in a sample that can be detected with better than a certain percentage counting accuracy, for example 5% accuracy. The MDA parameter is often used in order to set a detection limit for a detector system to assure that the MDA is measurable while monitoring a suspected radioactive contamination area (Bq/m²). MDA is also important in order to determine the ability of nuclear medicine equipments to measure lower levels of activities, mainly liquid scintillation counters or NaI(Tl)-well-counters. The parameter may also be used to determine detection limits for noninvasive equipments, such as scintillation cameras and PET scanners, keeping the administered activity low enough without impairment of the detector signal and diagnostic accuracy.

Different algorithms have been developed for the determination of MDA for different equipments and situations which can be found elsewhere.

**Related Articles:** Quality control, Activity, Optimisation


Minimum target absorbed dose

*Radiotherapy* The International Commission on Radiation Units and Measurements (ICRU) recommends a common method of dose specification which could be generally adopted to permit a comparison between the treatment practices. Normally a non-uniform dose distribution is obtained in the target volume and therefore for practical reasons it is useful to report specific doses as the minimum target absorbed dose.

The minimum target absorbed dose is the lowest absorbed dose in the target area. No area limit is recommended when reporting minimum target absorbed dose.

**Related Articles:** Mean target absorbed dose, Maximum target absorbed dose, Modal target absorbed dose, Median target absorbed dose, Hot spots

**Further Reading:** ICRU (International Commission on Radiation Units and Measurements). 1978. Dose specification for reporting external beam therapy with photons and electrons. ICRU Report 29, Washington, DC.

MIP (maximum intensity projection)

*Magnetic Resonance* See Maximum (minimum) intensity projection (MIP)

MIRD Committee

*Nuclear Medicine* The Medical Internal Radiation Dose (MIRD) Committee is a committee sanctioned by the Society of Nuclear Medicine (SNM) that provides fundamental quantities used for risk assessment, radiation dosimetry/protection and radiation therapy. The MIRD Committee develops methods, models, assumptions and mathematical models for calculation of the internal radiation dose.

MIRD formalism

*Nuclear Medicine* The MIRD method is used to calculate the total absorbed dose to patients from nuclear medicine examinations. The method is named after the Medical Internal Radiation Dose Committee of the Society of Nuclear Medicine.

With the *mean absorbed dose per cumulated activity* $S$, it is possible to calculate the radiation dose to target organs from radioactivity accumulated in one or several source organs. The source and the target organ can be the same organ; in fact often the major part of the absorbed dose originates from radioactivity within the target organ itself, often referred to as self dose. Other organs are considered to be source organs if they contain a concentration of radioactivity higher than the average concentration of radioactivity in the body.

The radiation dose to a target organ is determined in three steps:

1. *Cumulated activity*, $\bar{A}$: The first step is to determine the kinetics of the tracer, that is basically the time spent by the radioactivity in a specific organ and also the amount of activity present in each source organ. This is usually done with a gamma camera, but can involve other methods such as blood, urine and faeces samples. The activity concentration in an organ will show a temporal variance, that is from initial uptake to total clearance, and the behaviour is modelled by a *time-activity function* which differs between organs. The *cumulated activity* is the integration of the *time-activity function*, i.e. the total numbers of disintegrations in the source organ.

2. *Equilibrium absorbed dose constant*, $\Delta$: Secondly the total amount of emitted energy from the radioactivity in the source organ is calculated. Decisive parameters are the energy of the emitted radiation and radiation type.

3. *Absorbed fraction*, $\Psi$: The fraction of the emitted energy source organ absorbed in the target organ is determined. The dose contribution depends on the anatomic relationship between the target and source organ (size, shape and distance between the two), radionuclide characteristics (energy, particle or non-particle radiation, etc.) and the material composition of the tissue.

To calculate the radiation dose, $\bar{A}$ and $\Psi$ must be calculated for each emission, $i$. The radiation dose procedure is somewhat tedious, especially when dealing with different emission types and many source target pairs. To simplify the calculation procedure, the *mean dose per cumulated activity* $\bar{S}$ is introduced. $S$ is calculated for each radionuclide and for individual source–target pair. The total radiation dose received by a target organ from a specific source organ is

$$D(t_i \leftarrow r_j) = \bar{A} \times \frac{1}{m_i} \sum_i \phi_i (r_i \leftarrow r_j) \Delta = \bar{A} \times S(r_i \leftarrow r_j) \quad (M.9)$$

where $D(r_i \leftarrow r_j)$ is the radiation dose received in target organ $r_j$ from source organ $r_i$; $m_i$ is the target organ mass. More information about the different steps in the radiation dose calculation can be found in their respective articles (see Further Readings).

To create patient-specific S-factors, a set of 3D CT slices are required. From these, the segmentation of source and target organs is performed. This is then run through a Monte Carlo code to generate a table of $S$-factors. This process is so far too tedious to use in the clinic, so a set of $S$-factors derived from standard phantoms are almost always used. In this step, assumptions are made about the human anatomy, namely that all patients share the same internal structure, when in fact the structure varies from patient to patient, especially patients with pathological abnormalities. $S$-factors can be found from MIRD, OLINDA, ICRP.

The models (or phantoms) used to model the human body were published by the MIRD Committee of the Society of Nuclear
Misregistration  

Related Articles: Cumulated activity, Equilibrium absorbed dose constant, Absorbed fraction, Mean dose per cumulated activity, Limitations to the MIRD formalism  


Mixed radionuclides  

(Urology) The term mixed radionuclides refer to the use of multiple radioisotopes in nuclear medicine. Different imaging radionuclides (i.e. gamma emitters) are labelled to two different tracers in order to study two separate biological processes that spatially overlap. The two tracers can be spatially separated using an energy separation. Another application involves the use of one imaging radionuclide and one radionuclide with a therapeutic purpose labelled to the same tracer. The imaging radionuclide allows the user to perform in vivo evaluation of the targeting to tissue ratio.

Mixing time (*T*<sub>m</sub>)  

(Magnetic Resonance) Mixing time is the term used to refer to the interval between the second and third 90° pulses in the STEAM pulse sequence. During this interval, magnetisation is stored along the z-axis and undergoes *T*<sub>r</sub>-relaxation, introducing a weighting exp(−*T*<sub>m</sub>*T*<sub>r</sub>) into the final signal. This weighting is rarely significant, and there is a more significant reduction in *T*<sub>r</sub>-weighting due to storage of the magnetisation in the z direction, yielding an effective reduction in TE. Movement, including diffusion, during *T*<sub>m</sub> is a known problem with STEAM. Physiological motion during this interval can result in further loss of signal from the VOI, over and above the inherent 50%. A variety of approaches exist to compensate for this.

The development of multiple quantum coherences during *T*<sub>m</sub> results in complex behaviour of *J*-coupled species during the STEAM sequence.  

Related Articles: Magnetic coupling, STEAM  


Mobile shield  

(Radiation Protection) A mobile shield is a protective shield used for the protection of the worker during the use of ionising radiation.  

Fluoroscopy and Interventional Applications: Additional protective mobile shields should be available in fluoroscopy and in interventional radiology rooms, which include  

1. Ceiling suspended protective screens  
2. Protective lead curtains mounted on patient table  
3. Protective lead curtains for the operator if the x-ray tube is placed in an over the couch geometry and if the radiologist must stand near the patient

Mobile X-Ray Equipment: Mobile shield, with generally 2.5 mm Pb equivalent, is used to protect the operator when mobile x-ray equipment is used to perform x-ray investigation, in the ward, at the patient’s bed. The mobile shield will have a glass window with the same amount of protection (lead equivalent) in order to allow the operator to see the patient while performing the investigation.  

Mammography: Usually light, transparent shields are incorporated with the equipment in order to allow the operator to see the patient and remain in the room during the exposure.  


Mobile target volume  

(Radiotherapy) The mobile target volume has been suggested as an extension to the ICRU 50 formalism (gross tumour volume, planning target volume, clinical target volume, etc.). The mobile target volume would include an additional ‘safety’ margin to allow for subject motion such as respiration.  

Related Articles: Gross tumour volume, Planning target volume, Clinical target volume  


Mobile unit  

(Diagnostic Radiology) Mobile unit is the medical jargon used for any type of small low-power x-ray equipment (radiographic or fluoroscopic) on wheels. Mobile units are powered mainly by high frequency generators and use low-power x-ray tubes. Some radiographic mobile units use capacity discharge generators or monoblock generators. Mobile units are convenient to move into patient rooms or the operational theatre, as well to be used as field units in military hospitals. All mobile units are powered either by a battery or directly from the mains.
Modal target absorbed dose

*(Radiotherapy)* The International Commission on Radiation Units and Measurements (ICRU) recommends a common method of dose specification which could be generally adopted to permit a comparison between treatment practices. Normally a non-uniform dose distribution is obtained in the target volume and therefore for practical reasons it is useful to report specific doses such as the modal target absorbed dose.

The modal target absorbed dose is the absorbed dose that occurs most frequently at calculation grid points in the target area.

The modal target absorbed dose depends on the dose calculation method and exceptionally more than one modal target absorbed dose can be found in a particular patient.

**Related Articles:** Mean target absorbed dose, Minimum target absorbed dose, Median target absorbed dose, Hot spots

**Further Reading:** ICRU (International Commission on Radiation Units and Measurements). 1978. Dose specification for reporting external beam therapy with photons and electrons. ICRU Report 29, Washington, DC.

Modalities

*(General)* A modality refers to a family of imaging systems based on the same physical phenomena, for example SPECT and PET. Modality also refers to an individual imaging system in dual or multimodality system, for example the SPECT in a SPECT-CT system.

Mode conversion

*(Ultrasound)* When a longitudinal wave crosses the boundary (at an angle) into a solid material, some of the energy can cause particle movement in the transverse direction. This is called mode conversion.

**Related Articles:** Transversal wave, Longitudinal wave

Modulation transfer function

*(Diagnostic Radiology)* Modulation transfer function (MTF) represents the response of an imaging system to an input signal of continuously varying spatial frequency. Modulation of a signal is defined as

\[
\text{Modulation (M)} = \frac{I(\text{max}) - I(\text{min})}{I(\text{max}) + I(\text{min})}
\]

where

- \(I(\text{max})\) is the maximum intensity of the signal (e.g. white in greyscale image)
- \(I(\text{min})\) is the minimum intensity (e.g. black in greyscale image)

This can also be interpreted as the total contrast of the image. The important radiological issue is what a radiologist will be able to distinguish in the images (see the following hyperlink for a discussion of issues), but the MTF is also reported for standard digital cameras. MTF is defined as the ratio between the modulation of the resulting image \(M(i)\) and the modulation of the input object \(M(o)\) as function of spatial frequency \(\nu\):

\[
\text{MTF}(\nu) = \frac{M(i)}{M(o)}
\]

In the determination of the performances of medical imaging systems, MTF gives a complete description of the spatial resolution in the spatial frequency domain. Two main procedures can be followed for the evaluation of MTF.

**Bar Test Object:** These are built to represent an object with groups of black and white line pairs, giving an input image with increasing spatial frequency. Each group of line pair constitutes a specific spatial frequency, expressed as line pair per length unit (e.g. lp/mm), and the measurement of the modulation of each group represents a set of experimental points for the determination of the MTF curve. MTF is expressed as per cent variation in the image modulation normalised to the modulation at low frequency where there is no significant loss of the signal.

**Fourier Transformation:** Spatial resolution is related to the capability of an imaging system to represent the sharpness of a high contrast edge. Starting from the experimental measurement of the edge response function (ERF) of a high contrast edge image it is possible to evaluate, by derivation, the line spread function (LSF) which represents the response of the system to a high contrast linear signal (Dirac delta function). The Fourier transformation of the LSF represents the MTF of the image system in the spatial frequency domain.

An image has two essential components: edges where intensities (densities, grey values) change, and contrast, the amount by which they change. A sharp edge is a line along which change is steep. Imperfect imaging systems will affect such lines. They may spread the region on which the change happens, or they may reduce size of the change, and both effects amount to the same up when the image is normalised. This duality explains how contrast (or modulation if normalised) and edge sharpness relate. To describe sharp edges in the frequency domain, we need high spatial frequencies. The MTF, also called spatial frequency response, describes how the system affects different spatial frequencies differently (Figure M.37).

**Abbreviation:** MTF = Modulation transfer function.

**Related Articles:** Point-spread function, Optical transfer function


Modulation transfer function (MTF)

*(Magnetic Resonance)* Modulation transfer function is a means of quantifying the spatial resolution of an MRI system. MTF specifically looks at perceived contrast within the image. To measure MTF, a test object with a high contrast, angled block is scanned (Figure M.38). An edge response function can then be obtained from the image. By differentiating the edge response function, a line spread function is produced. The full width half maximum measure is taken from the line spread function to give a good indication of the spatial resolution.

**Modulator** *(Radiotherapy)* The modulator in a linac generates high voltage (approximately 100kV) and current in short pulses (the order of microseconds) to the radiofrequency power source for the production of the microwave radiation (either klystron or magnetron) and to the electron gun.

Typically the modulator circuitry is kept in a separate cabinet which can be located in the treatment room, or in a control room nearby, and with the high voltages involved interlocks on the door of the modulator cabinet are used.

More detail on the circuitry used and information on the PFN and PRF are given in Further Reading.

**Abbreviations:** PFN = Pulse forming network and PRF = Pulse repetition frequency.
Molecular excitation
(Nuclear Medicine) This is a process in which a molecule is excited from a low energy state to high energy state. The energy state of a molecule is modelled by the molecular Hamiltonian where each solution or eigenvalue represents an energy state that is the sum of electronic, rotational, nuclear and translational components.

Molecular mass
(Nuclear Medicine) The molecular mass is the weight of one molecule expressed as a multiple of 1/12 of the total weight of a $^{12}$C atom. Molecular mass is sometime referred to as molecular weight. The molecular mass of a single molecule is not always identical to the average molecular mass in a sample. The molecular mass differs because of the natural occurrence of isotopes.

Molecular weight
(Nuclear Medicine) See Molecular mass

Molecule
(General) A molecule is an electrically neutral group of two or more atoms connected by covalent bonds. Atoms attached by non-covalent bonds, such as hydrogen bonds or ionic bonds, are not strictly regarded as molecules. However, the term may sometimes be used to describe certain charged organic molecules or biomolecules. A molecule may contain atoms of the same chemical element (e.g. oxygen gas O$_2$) or different elements (e.g. water H$_2$O).

Molecules cannot be defined for ionic crystals, covalent crystal networks or metallic substances, although these normally consist of repeating units. Glasses contain atoms held together by chemical bonds but they are amorphous meaning that a molecule, or even a regular repeating unit, cannot be defined.

Related Articles: Atom, Molecular mass

Molière scattering theory
(Radiation Protection) The Molière scattering theory attempts to describe the small-angular distribution of electrons that have undergone elastic collisions with the atomic nuclei of an absorbing
medium – that is, the electrons do not lose energy during these scattering interactions, but do change direction.

This theory, together with the Klein–Nishina differential cross-section, Bethe–Bloch equation and others, attempts to describe interactions between ionising radiation and matter at an atomic level, and forms the mathematical basis for radiation dosimetry based on Monte Carlo statistical modelling.

Related Articles: Elastic scatter, Klein–Nishina differential cross-section, Bethe–Bloch equation

**Molybdenum**

**(General)**

| Symbol | Mo |
| Element category | Transition metal |
| Mass number A of stable isotopes | 92 (14.84%); 94 (9.25%); 95 (15.92%); 96 (16.86%); 97 (9.55%); 98 (24.13%); and 100 (9.63%) |
| Atomic number Z | 42 |
| Atomic weight | 95.94 kg/kg-atom |
| Electronic configuration | 1s² 2s² 2p⁶ 3s² 3p⁶ 3d¹⁰ 4s² 4p⁶ 4d⁵ 5s¹ |
| Melting point | 2896 K |
| Boiling point | 4912 K |
| Density near room temperature | 10,280 kg/m³ |

**History:** During the two world wars, molybdenum alloys were commonly employed in the armour plating of tanks. In modern times molybdenum compounds are used in pigments, catalysts and electrodes. Molybdenum is also used as target material in the production of some x-ray tubes.

**Isotopes of Molybdenum:** Molybdenum has 35 known isotopes, 7 of them are stable. The isotope of interest in medical physics is ⁹⁹Mo produced either as a product of uranium-235 fission or as neutron activation product. Whereby, stable molybdenum-98 is bombarded with thermal neutrons in a nuclear reactor.

| Isotope of molybdenum | ⁹⁹Mo |
| Half-life | 65.94 hours |
| Mode of decay | β⁻, γ |
| Maximum decay energy, E_max | 1.214 MeV, γ: 0.74 |

**Medical Applications:** Technetium generator: The daughter product of ⁹⁹Mo is ⁹⁹ᵐTc, a radionuclide widely used in diagnostic nuclear medicine studies (6.01 h half-life). To ensure that technetium radiopharmaceuticals are readily available, many clinical nuclear medicine departments run molybdenum-based technetium generators on site (in hospital radioopharmacies).

**Related Articles:** Technetium-99m, Technetium generator, Target of x-ray tube

**Molybdenum breakthrough**

*(Nuclear Medicine)* Molybdenum breakthrough occurs when the radioactive parent ⁹⁹Mo is partially eluted with the daughter ⁹⁹ᵐTc from a radionuclide generator. Samples used for clinical imaging must not contain any ⁹⁹Mo because it will contribute to an unnecessary radiation dose to the patient. One can estimate the ⁹⁹Mo contamination by using specific detector arrangement with a thick photon absorber (e.g. 30 mm of lead) between the crystal and the sample. The 140 keV photons emitted from ⁹⁹ᵐTc will be attenuated by the shield. But for the 740–780 keV photons from ⁹⁹Mo the absorber is relatively transparent.

**Monitor chamber** *(Radiotherapy)* A monitor chamber is included in the head of a linear accelerator to control continuously the radiation beam. The monitor chamber consists of a set of flat transmission ionisation chambers which are usually mounted close to the accelerator window between the flattening filter and the collimators so that the beam passes through them. The monitor chamber must cause minimal perturbation of the radiation beam and therefore the electrodes are formed by deposition of carbon or metal onto a thin substrate of mica or a plastic foil such as Melinex or Kapton. The substrate has a typical thickness of 0.1 mm. The reduced thickness of the electrodes permits a reduction of the radiation scattering, the bremsstrahlung contamination of the beams, the broadening of the electron energy spectrum and the production of low energy secondary electrons. The collection efficiency of the monitor chamber must be about 99% at the high dose rate of the linear accelerator and this requires a polarising voltage of about 300 V.

The monitor chamber consists of multiple parallel electrodes which form a double ionisation chamber system for two independent dose monitoring channels. The chamber plates are mounted so that one is rotated 90° on the beam axis from the other. This arrangement allows the on-line testing of the beam symmetry and flatness in the beam’s radial and transverse planes, respectively. Beam flatness and symmetry are controlled by feedback circuits that run from the ion chambers to the bending magnet’s beam steering coils.

The response of the monitor chamber depends on the mass of gas between its electrodes and therefore it is necessary to correct the system response for any variation of temperature, humidity and pressure in the chamber volume in case of an unsealed monitoring chamber filled with air. Sealed chambers have been also used. The response of the monitor chamber is influenced by the backscatter radiation from the jaws and is dependent on the radiation energy. The prescribed dose needs to be delivered reproducibly for each patient treatment. To achieve this routinely, one set of electrodes of the monitor chambers is used to monitor the beam output, so that the dose can be measured and delivered reproducibly.

The units of the signal recorded by this set of electrodes are referred to as monitor units (MU). One MU of dose has been delivered when the monitor chambers have detected a preset dose. The ionisation current I, is proportional to N, number of radiation quanta per unit of time passing through the chamber and to Np, number of radiation quanta incident on the phantom. The absorbed dose rate at a reference point in the phantom is also proportional to Np.

**Monitor unit** *(Radiotherapy)* Monitor units (MUs) are used in radiotherapy to measure out the dose delivered by a linear accelerator. They are the equivalent of ‘beam-on’ time for a Cobalt-60 machine. Commonly, the linear accelerator is calibrated to deliver 1cGy/MU at reference conditions for photon beams and at dmax for a 10 cm × 10 cm field at SSD = 100 cm for electron beams.

MUs are measured by the pair of transmission monitor ionisation chambers mounted in the treatment head of the linear accelerator. The dose monitor chambers are arranged either as a
Mono-energetic beams of ionising radiation are very difficult to obtain. They would represent a great advantage for medical therapy as well as diagnosis with ionising radiation. The development of synchrotrons and other advanced equipment as laser plasma accelerators make it possible to select monochromatic beams; but this kind of equipment is still far from the routine medical practice.

**Monte Carlo calculations**

 Nghệ thuật Monte Carlo là một phương pháp simulación số học được sử dụng để giải quyết vấn đề chứa các sự kiện ngẫu nhiên được mô phỏng. Các quá trình vật lý, như ánh sáng và electron tương tác, có thể được mô phỏng để tiếp tục có thể thực hiện được sự biến đổi của chất lượng phóng xạ. Với các yếu tố cần thiết của một hệ thống Monte Carlo, các simulação được thực hiện để tìm ra các kết quả mong muốn.

**Related Article:** Acquision time


**Monte Carlo method**

**Radiotherapy**

The Monte Carlo method involves the use of random number based statistical sampling to model a series of physical processes. It is particularly useful for modelling a series of physically independent events.

A particularly common application of Monte Carlo in radiology and radiotherapy is the modelling of ionising radiation transport. The concept is that the ionising radiation is described by modelling many independent samples and the statistics of the samples is analysed to determine the cumulative behaviour of many particles of the radiation. For each particle, the energy, direction, etc. are randomly sampled at the start of the particle’s ‘history’ (where history is used to describe the chain of events that happens to each individual particle that is modelled). The interaction of the radiation with matter is then randomly sampled according to the probability of interaction and the nature of that interaction. Any secondary radiation that is created from the interactions is then followed similarly, in order to describe the entire radiation effect. This methodology may be used to model the dose delivered to the patient by a treatment (such as external beam radiotherapy or brachytherapy), to determine the dose required to form an x-ray image, or to calculate the dose response of a detector such as an ionisation chamber.

A series of off-the-shelf packages are often used for Monte Carlo radiation transport, these include EGS/BEAM (where EGS stands for electron gamma shower), GEANT, PENELPO and others. Often the packages require a user file to specify the geometry and physical requirements of the problem to be solved.

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**Abbreviation:** MU = Monitor unit.

**Abbreviation:** mAb = Monoclonal antibody.

**Related Articles:**

- **Monoclonal antibodies (mAb)**
  - (Nuclear Medicine) Antibodies are the body’s natural defence against foreign substances known as antigens (e.g. disease-causing bacteria and viruses). Each antibody has an affinity for a specific antigen, and some antibodies, once activated, will continue to provide resistance (this is the basis for the production of vaccines).

- **Related Articles:**
  - High voltage generator, Voltage waveform

- **Further Readings:**

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**Mono-block generator**

(Mono-block generator)

Although called mono-block HVG, this is not a new design of HVG, but is a construction of the x-ray tube housing, which incorporates not only the x-ray tube, but also the high voltage transformer, the rectifiers, etc. All components are immersed in isolation oil in the mono-block box. Most often these units are used for dental radiography, having a stationary anode tube and two pulse rectifier. This design can also be used with low-power rotating anode tube in some mobile x-ray equipment. Mono-block generator is also called single block generator.

**Related Articles:**

- High voltage generator, Voltage waveform

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**Monoclonal antibodies (mAb)**

(Nuclear Medicine) Antibodies are the body’s natural defence against foreign substances known as antigens (e.g. disease-causing bacteria and viruses). Each antibody has an affinity for a specific antigen, and some antibodies, once activated, will continue to provide resistance (this is the basis for the production of vaccines).

In the past antibodies have been produced by injecting a laboratory animal with an antigen and when antibodies have been formed, collecting these from the blood serum. One problem with this method is that it produces a very small quantity of usable antibody. Monoclonal antibody technology on the other hand provides a technique to produce large quantities of pure antibodies. This is achieved by forming a hybrid of cells that produce antibodies naturally together with cells that can grow continually in cell-culture. The result of this cell fusion is called a hybridoma and it will continually produce antibodies. These antibodies are called monoclonal antibodies (abbreviated as mAb) because they come from only one type of cell (the hybridoma cell). Antibodies produced in the conventional manner are called polyclonal because they come from substances that contain many kinds of cells.

The technique for producing monoclonal antibodies was first described by Kohler and Milstein (1975). They found that when myeloma cells were fused with antibody-producing mammalian spleen cells, the resulting hybridomas produced large amounts of monoclonal antibodies.

In nucleic medicine, antibodies against tumour specific antigens are used for tumour targeting. They are then labelled with a radionuclide and can be used for diagnostics (radioimmunodiagnostics) or therapy (radioimmunotherapy).

**Abbreviation:** mAb = Monoclonal antibody.

**Related Articles:**

- Radioimmunotherapy, Tumour targeting

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**Mono-energetic beam**

(Radiation Protection) Mono-energetic beams have a single value of linear energy transfer (LET). The half value layer (HVL) thickness, (namely the thickness of a specific material which reduces the intensity of the radiation entering the material to half) will remain the same. Therefore the 1st HVL is equal to the 2nd HVL. The homogeneity factor (HF) is the ratio between the 1st and the 2nd HVL and is a measure of the polychromaticity of the beam. In case of mono-energetic beams, HF is equal to one.
Monte Carlo photon transport simulation

Statistical Sampling: A variety of approaches are available for statistically sampling events to determine a particle's history. An example is the cumulative probability distribution, CPD. In this approach, if the probability is \( p(x) \) as a function of the independent variable, \( x \), then the summed probability between \(-\infty \) and \( x \):

\[
P(x) = \int_{-\infty}^{x} p(y)dy
\]

is calculated. A random number, \( R \), between \( 0 \) and \( 1 \) is generated and the value of \( x \) for which \( P(x) = R \) is used to find \( x \) by inversion.

To illustrate this, consider the following. Figure M.39 shows a hypothetical Gaussian angular distribution for scattering of a particle. Figure M.40 shows the cumulative probability distribution. We may sample this distribution in 1 by generating a random number between 0 and 1 (\( R \) in Figure M.40). We then read off the value of the angle for which the CPD is equal to \( R \). This gives a value of \( \theta \).

Abbreviations: CPD = Cumulative probability distribution and EGS = Electron gamma shower.

Related Articles: Radiotherapy treatment planning, X-rays, Electrons


Monte Carlo photon transport simulation

(Nuclear Medicine) The Monte Carlo method is useful for solving very complex problems of a stochastic nature. In nuclear medicine, radiation is used for imaging and for the treatment of cancer. In imaging, it is mainly the photon transport that is used to create images of radionuclide distributions. The Monte Carlo simulation of a photon transport includes simulation of photo-absorption, Compton scattering and coherent scattering. By sampling from probability distribution functions, each individual photon can be followed in a phantom towards a simulated camera. It is necessary to estimate for each interaction, the reduction in energy as well as the scattering angles. The simulation needs to include a large number of histories in order to maintain good statistics in the results.

Examples of Monte Carlo codes for photon transport simulation in nuclear medicine applications are the SIMIND code, SIMSET code and the GATE/GEANT4 code. More information about these codes can be found on the Internet.

Most probable energy

(Radiotherapy) In a linac the electron beam in the accelerating component reaches a specific final energy that depends on the design of the accelerating structure. The intrinsic electron beams are almost mono-energetic because their energy distributions are very narrow. After leaving the accelerating structure the spectrum degrades and the clinical electron beams are characterised by a number of parameters such as the maximum energy \( E_{\text{max}} \), the most probable energy \( E_p \) and the mean energy \( E \). In Figure M.41, the typical distribution of the electron beam energy is shown before leaving the accelerating structure of the linac [graph (a)]; at the phantom surface [graph (0)]; and at a depth \( z \) in the phantom [graph (z)].

At the phantom surface two energy parameters are of particular interest: the most probable energy \( E_{p,0} \) that corresponds to the peak of the distribution and the mean energy \( E_0 \). These parameters can be easily determined by practical measurements and are utilised to specify the electron beam quality.

The most probable energy of the electron beam at the surface of the phantom \( E_{p,0} \) is related to the practical range in water \( R_p \), expressed in centimetres, by

\[
E_{p,0} = 0.22 + 1.98R_p + 0.0025R_p^2 \quad (\text{MeV})
\]

The energy losses of the electron beam when it traverses layers of matter results in a broader spectrum. Experimental data indicate that there is approximately a linear relationship between most probable energy \( E_{p,0} \) on the phantom surface and at the depth \( z \) and is given by

\[
E_{p,z} = E_{p,0} \left( 1 - \frac{z}{R_p} \right)
\]

where

- \( E_{p,0} \) is the most probable energy at the surface
- \( R_p \) is the practical range of the electron beam
- \( z \) is the depth in the phantom

Related Articles: Mean electron energy, Electron practical range

Motion artefact

(Diagnostic Radiology) Motion artefacts can be the result of voluntary (swallowing, breathing), or involuntary (cardiac motion,
Motion artefacts

(Physiological movement) patient movement. These artefacts are often seen in CT scanning. Motion during the course of a scan results in streaks in the scan plane images. This is due to inconsistencies in the acquired projection data. Movement in between rotations results in misregistration artefacts in multiplanar reconstructions (MPRs) and 3D reconstructions.

During the course of a CT scan, patients are usually requested to hold their breath in order to eliminate breathing artefacts. They are given positioning aids, and in some cases may be immobilised. Young children or uncooperative patients may need to be scanned under a general anaesthetic.

Motion artefacts can be corrected with the use of software which reduces the weighting of views at the beginning and end of a rotation (Figure M.42). However, modern, multislice CT scanners have much reduced examination times and therefore motion artefacts are less of an issue.

Motion artefacts are still a particular problem in cardiac scanning due to the rapid movement of the heart. Fast rotations, of less than 300 ms, are now available on the latest CT scanner models. Images are reconstructed from partial scan data to improve temporal resolution still further, and ECG gating techniques are used to select the cardiac phase of least motion for reconstruction. Dual source CT scanners are particularly suited to imaging the heart because of the improvement in temporal resolution resulting from acquiring data simultaneously with two x-ray tubes.

Related Articles: Artefact, Beam hardening, Cone beam artefact, Dual source CT, Helical artefact, Image artefact, Metal artefact, Multislice CT scanner, Partial volume effect (artefact), Ring artefact

Motion artefacts

(Magnetic Resonance) Motion artefacts are caused by movement of the patient. These movements can be divided into two categories, physiological motion such as breathing and bowel motion, and gross patient movement such as small twitches. The artefact typically appears as a ghost along the phase-encoding direction.

Flow artefacts are another type of motion artefact caused by the flow of blood or cerebrospinal fluid within the body. Due to the flow of the protons, mismapping on the image is caused leading to ghosting. Flow compensation can be used to reduce this artefact.

Physiological motion artefacts can be reduced by respiratory and cardiac triggering, the use of breath-hold pulse sequences, or presaturation pulses, depending on their origin.

General patient movement cannot be avoided, but can be reduced by ensuring the patient knows they must stay as still as possible during a scan and by reducing scan times as much as is acceptable.

Abbreviation: SNR = Signal to noise ratio.

Related Articles: Ghost artefact, Periodic motion

Motion unsharpness

(Diagnostic Radiology) Image unsharpness due to movement of patient during the x-ray exposure. Typical motion unsharpness is
observed in chest imaging when the patient breathes during the radiography (or CT scan).

**Mottle, quantum**
*Diagnostic Radiology* Quantum mottle refers to the pattern of quantum noise, or random variations, in a radiographic image which is due to the statistical fluctuation of x-ray photon absorption and consequent light photon emission by an intensifying screen or digital radiology scintillator screen. The faster the intensifying screen, or the higher the kV, the more light photons are produced and so fewer x-ray photons actually contribute to a final image of a desired optical density or pixel value.

Since the emission and detection of photons are normally distributed, the noise, or randomness, associated with the number of photons is proportional to the square root of the number of photons. So with fewer x-ray photons contributing to the image, the greater is the noise, or mottle, seen in the image.

**Moving grid**
*Diagnostic Radiology* The moving antiscattering grid is invented by Hollis Potter, later named Bucky–Potter, but now referred to as only Bucky grid. The advantage of the moving grid is that the lines of the grid are not seen on the image (a problem seen in images with stationary grids, invented by Bucky).

The moving grid is initially displaced from the centre of the film (i.e. is not centred). It starts its slow motion with the beginning of the exposure and ends it at the end of the exposure, passing through the centre at approximately half-time of the exposure. Some moving grids do have such one-directional movement, but vibrate during the exposure.

As a result of this movement, the pattern of the led lamellas of the grid is blurred over the image. However this movement reduces not only the secondary (scatter) radiation, but also the primary radiation. To compensate this, exposure factors (i.e. patient dose) have to be increased by some 10%–15%.

**Related Articles:** Grid, Bucky

**MPD (maximum permissible dose)**
*Radiation Protection* See Maximum permissible dose (MPD)

**MPR (multiplanar reconstruction)**
*Magnetic Resonance* See Multiplanar reconstruction (MPR)

**MRA (MR angiography)**
*Magnetic Resonance* See Magnetic resonance angiography (MRA)

**MR microscopy**
*Magnetic Resonance* The spatial resolution available to conventional clinical magnetic resonance imaging (MRI) is limited by the amplitude of the switched magnetic field gradients and by the static magnetic field strength of the imaging system, which limits the signal to noise (SNR) achievable in an acceptable examination time. This combination of factors limits resolution to around 1 mm, or a little better with specialist sequences and/or equipment (typically 0.5 mm or above).

However, outside the clinical arena, very high field strength (>10 T), small bore MRI systems are available with high gradient field strengths, and in many non-clinical applications an imaging time of several hours is perfectly acceptable. This enables MR microscopy – usually defined as imaging with a spatial resolution of better that 100 μm.

The limit of resolution available to conventional MR microscopy is about 1 μm. Thus the technique does not compete with other microscopy techniques, but rather complements them because of the capability of MRI to produce images dependent on a wide range of nuclear, physiological and chemical parameters.

Applications of MR microscopy have included botany, materials science and imaging of porous media. It is also possible to image animal models of relevance to medicine non-destructively, which has considerable potential for developmental biology and genetics (Figure M.43).

Recently, alternative approaches to MR microscopy have pushed the resolution limit further. Magnetic resonance force microscopy (MRFM) uses a sensitive cantilever to sense the force between a magnetic tip and individual spins within the sample. Resolution of around 90 nm has been achieved in this way.


**MR safe**
*Magnetic Resonance* The term MR safe indicates that the device has been evaluated to show that when used in an MR environment, its use does not present additional risk to the patient. However MR Safe does not assure that its use does not affect the quality of the diagnostic information. The term MR safe without specification of the MR environment to which the device has been tested should be avoided to prevent any incorrect interpretation.

**MR safety (new definitions)**
*Magnetic Resonance* The ASTM (2005) defines three classifications of devices/objects from an MRI safety perspective:

- **MR safe**: an item containing no metallic parts and posing no known hazard in all MRI environments.
- **MR compatible**: an item demonstrated to pose no known hazard in a specified MR environment. The ‘specified environment’ defines the field strength, maximum gradient rate of change, SAR or other conditions under which the item has been tested.
- **MR unsafe**: an item known to demonstrate hazards in all MRI environments.

These definitions have been adopted by the American College of Radiologists and by the MHRA in the United Kingdom. The definitions are designed to avoid the imprecise use of the terms ‘MR compatible’ and ‘MR safe’. In particular the definitions emphasise that the risk associated with a given device depends on the MRI environment in which it is used: For example, a device may be safe at 1.5 T but not at 3 T. ‘MR safe’ is a term appropriate only for non-metallic items, safe under all conditions. Most devices designed for use in an MRI environment will be ‘MR compatible’, that is safe under defined conditions of use. The conditions under which such devices are compatible must be stated on documentation accompanying the device.

Devices used in the MRI environment should have labels affixed following the convention shown in Figure M.44.

MRI safety

(Magnetic Resonance) MRI safety is concerned with the good management of MRI units and MRI examinations in order to reduce the risks associated with MRI scanning. Guidance on the safety issues to be considered in MRI is available from a number of sources including the ACR and the MHRA. The main safety concerns are as follows:

1. Projectile effect (see Projectile effect).
2. Peripheral nerve stimulation (see Peripheral nerve stimulation).
3. Incompatible implants: Incompatible passive implants (e.g. ferromagnetic aneurysm clips) risk moving due to the strong static field of the scanner, with the potential...
MRI safety begins at system commissioning, where room layout should be designed to minimise the possibility of unauthorised access to the scan room. Warning signage must be provided on doors leading to the unit. Rooms should incorporate oxygen alarms to monitor oxygen levels. In the event of a low oxygen alarm, the alarm may be used to trigger an extract fan to remove helium from the room.

MRI systems on the market must conform to IEC requirements. The IEC defines three levels of MRI operation: normal mode, first level and second level (or experimental mode). The maximum static field, SAR and gradient field switching rates for each of these levels are defined.

Normal mode operation will not result in whole body temperature rises of greater than 0.5°C and carries a low risk of peripheral nerve stimulation. In first level mode one or more of the MRI outputs may lead to physiological stress. Temperature rises are restricted to less than 1°C and there is an increased risk of Peripheral nerve stimulation. The MRI system will present a warning to the user where the parameters of the examination result in first level operation. Improved patient monitoring may be required in first level examinations and some patient groups (e.g. pregnant patients) may be restricted to normal mode exams.

On second level mode, one or more of the system outputs may result in a significant risk to the patient. Second level is for use only in research under strictly controlled conditions and with ethics committee approval. In routine use second level mode is not accessible to the system user.

MRI units will have written local rules to help reduce risks. The rules will address the following:

1. Definition of roles and responsibilities of personnel with respect to MRI safety.
2. Definition of the scan room and rooms accessing the scan room as controlled areas. The ACR set out a model, multi-level zonal definition for MRI suites, listing appropriate levels of vigilance for each zone. As a basic requirement, entry to the scan room must be strictly controlled and limited to screened patients and personnel.
3. Patient screening procedures: All patients must be screened to rule out the possibility of a contraindication to MRI. Patients are required to complete a questionnaire on their medical history prior to scanning. Those with implants incompatible with the MRI procedure or with metal fragments in their eyes will be excluded. Published data are available in commonly used implants and the conditions under which they are compatible. Exclusion criteria for the use of contrast agents based on current practice should be defined locally.
4. Screening of objects: No object of unknown MRI safety status may be allowed to enter the scan room. Items which are known to have no metallic components may be safely brought into the scan room. Items tested and defined as MRI Conditional may be used within the scan room under defined conditions (see MRI safety, new definitions).
5. Staff training: Staff must be familiar with the risks associated with MRI and with patient set up procedures to reduce the risk of RF burns and heating, the provision of hearing protection and with room evacuation procedures. Staff must also be familiar with the concepts of normal, first level and second level system operation.

Related Articles: MRI safety new definitions, Projectile effect, Peripheral nerve stimulation

MRI transparent

(Magnetic Resonance) A material or device is ‘MRI transparent’ if it does not generate detectable RF signal in an MRI scan. The transparent object will appear as a signal void in an MRI image. Plastics and other solid non-metallic objects that do not contain mobile protons are typically transparent to conventional clinical MRI systems.

MRS (magnetic resonance spectroscopy)

(Magnetic Resonance) See Magnetic resonance spectroscopy (MRS)

MRS voxel contamination

(Magnetic Resonance) The aim of single voxel localisation in MRS is to obtain a spectrum that originates from within a well-defined volume-of-interest (VOI) without contamination with signal from surrounding tissues. In practice, instrumental and pulse sequence limitations mean that there are a number of mechanisms whereby a spectrum may be contaminated with extraneous signal. Severe contamination can impact dramatically on the clinical value of a spectrum, and is usually neither immediately obvious from inspection of the spectrum, nor readily correctable.

Contamination may be broadly divided into ‘profile contamination’, due to imperfections in the sensitivity profile of the VOI, and ‘background contamination’, due to imperfect signal suppression outside the VOI.

Profile contamination arises from imperfections in the slice profiles generated by radiofrequency pulses in the MRS pulse sequence, together with other hardware and sequence imperfections and relaxation effects. The profile typically consists of a maximum sensitivity plateau at the centre of the VOI falling gradually to zero sensitivity at some distance outside the nominal VOI. On commercial MR systems, the definition of the VOI size relative to this profile, and hence the degree of contamination arising from signal within the profile but outside the nominal VOI, is a matter for the system manufacturer (Figure M.45).

Background contamination may occur in single-voxel MRS sequences in which the pulse sequence affects magnetisation in parts of the subject outside the VOI. Depending on pulse sequence details, and in some cases hardware factors, this may result in generation of contaminating signal from outside the VOI due, for example, to $T_1$ relaxation or imperfections in signal subtraction. Background contamination is often addressed by adding outer volume suppression (OVS) techniques to the pulse sequence.

Of the popular single voxel techniques, STEAM and PRESS, for proton MRS, are single-shot techniques with little potential for background contamination as long as the spoiler gradients are effective. Profile contamination is minimal, but tends to be worse with PRESS than with STEAM, as it is more difficult to design $180^\circ$ refocusing pulses with clean selection. ISIS, for phosphorus MRS, is prone to serious background contamination due to imperfect signal subtraction, exacerbated by a mechanism known as $T_1$ smearing (see article ISIS for details), but generally shows little profile contamination.

Related Articles: ISIS, PRESS, Single voxel spectroscopy, STEAM


ms selector

(Diagnostic Radiology) The ms (millisecond) selector is part of the controls of the HVG of an x-ray equipment. It controls the timer, allowing the radiographer to select the length of the exposure (in milliseconds). When this time elapses the timer interrupts the exposure.

Related Article: High voltage generator

MSAD

(Radiotherapy) See Multiple scan average dose (MSAD)
Multichannel analyser
(Nuclear Medicine) A multichannel analyser (MCA) is the standard device for collation and analysis of data from a radiation detector, for example scintillation or semiconductor detector. The MCA is used in either pulse height analysis (e.g. gamma spectroscopy) or multiscaler mode (time-activity diagram).

In the pulse height analysis mode, pulses from the detector are sorted and stored according to their amplitude (height), enabling the pulse height spectrum to be displayed. This is achieved by feeding the pulses into an analog-digital converter (ADC) which for each pulse generates a number (in binary form) proportional to the height of the pulse. This number is then used as the address of a memory location in RAM, so that each location or ‘channel’ records the number of pulses of a specific pulse height.

In the multiscaler mode of operation, the time spectrum of the pulses from a radiation detector is displayed, that is a time-activity plot.

In nuclear medicine the MCA is often used together with a NAI(Tl)-scintillation detector for in vivo measurements of the activity uptake in different organs, for example thyroid uptake measurements or as an intra-operative probe. It is also used for measurements of radioactive samples in vitro.

The MCA is also a valuable instrument together with a GeLi-semiconductor detector for quality control of radiopharmaceuticals, that is control of the radionuclide purity. Two examples of isotope impurities are $^{109}$In or $^{114}$In$^{m}$ in $^{111}$In-solutions and $^{202}$TI in $^{201}$TI.

The MCA is sometimes a built-in module in a detector system, for example scintillation cameras, or used as a stand-alone PC-based, software controlled PCI-bus MCA (Figure M.46).

Abbreviations: ACD = Analogue-digital converter, MCA = Multichannel analyser and SCA = Single channel analyser.

Related Articles: Analogue-digital converter, Single channel analyser, Quality control, Radionuclide purity


Multicrystal scanners
(Nuclear Medicine) In PET imaging the use of thin crystals improves the centre FOV spatial resolution. But the spatial resolution is degraded for peripheral sources because of crystal penetration by oblique incident photons. This effect is referred to as the depth of interaction or parallax effect. One common way to compensate for this effect is to use several layers of crystals instead of a homogenous crystal. The crystals in the different layers have different decay times and it is therefore possible to get the depth of interaction by analysing the pulse shape. It does not necessarily need to be different crystal types in each layer. For example, the decay time of Gd SiO:Ce (GSO) crystal can be altered by changing the amount of Ce (Cerium) in the crystal.


Multi-echo
(Magnetic Resonance) A multi-echo pulse sequence utilises several 180° RF pulses to recall more than one echo after a 90° RF excitation pulse. A common version recalls two echoes with two different echo times, that are collected using the same phase encoding and put into separate raw-data sets, hence providing two images of different contrast. A common combination is a pulse sequence collecting one proton-density-weighted image, requiring short echo time and long repetition time, and one $T_2$-weighted image requiring a long echo time and a long repetition time, that is a so-called Pd-$T_2$-protocol. The double contrast protocol can be based on either conventional spin echoes or the faster version, fast spin echoes.

Related Articles: Carr–Purcell (CP), Carr–Purcell–Meiboom–Gill (CPMG) Sequence, Fast spin echo, FSE, $T_2$-weighted

Multi-gated acquisition (MUGA)
(Nuclear Medicine) Multi-gated acquisition (MUGA) is a composite study of cardiac contractility and wall motion. Image acquisition is triggered in a position of interest in the cardiac cycle using the signal from an ECG (electrocardiogram). The examination is performed on patients with suspected coronary artery diseases or congestive heart failure. The test can be performed when the patient is at rest or exercising (called MUGA stress test). A stress test is usually performed on patients with possible coronary artery disease. One of the important parameters to monitor is the ejection fraction.

Further Reading: Gill (CPMG) Sequence, Fast spin echo, FSE, $T_2$-weighted

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Further Reading: Gill (CPMG) Sequence, Fast spin echo, FSE, $T_2$-weighted

Multi-gated acquisition (MUGA)
Multihole collimators for radioisotope scanners
*(Nuclear Medicine)*

A multihole collimator, often called multichannel, honeycomb or focusing collimator, is constructed of shielding material placed between the radiation detector and the gamma emitting radioactive source. Several holes through the shielding allow radiation to impinge on the detector from a specified spatial location only. The field of view of a multihole collimator is simply the superposition of all fields of view of each of the several holes at the focal plane they superimpose. Everywhere else the superposition is only partial, depending on the distance from the focal plane. Tapered holes must be used to make the maximum amount of crystal area available to the impinging photon fluence.

The spatial resolution is the ability to distinguish between two radioactive sources close together. The depth response depends on the focal distance. The sensitivity is its ability to detect radiation from the radioactive source.

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Related Articles: Radioisotope scanner, Collimator


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Multihole focused collimators
*(Nuclear Medicine)*

See Diverging collimator

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Multileaf collimator
*(Radiotherapy)*

Multileaf collimators (MLC) provide one of the most sophisticated collimation methods in radiotherapy, conforming the delivered beam closely to the target shape, thereby reducing the volume of normal tissue irradiated, and reducing the dose to critical structures near the boundary. They consist of 20–80 pairs of tungsten leaves arranged in two opposing banks, thick enough to provide similar attenuation to that of conventional secondary collimator jaws. Each leaf is normally 1-cm wide, and can move individually to create irregularly shaped fields, which conform closely to the shape of the projected planning target volume (PTV), with accuracy on the order of millimetre. Some MLCs have micro-MLCs within the centre of the FOV that allow for finer collimation (e.g. width 0.5 cm cf 1 cm). See Figure M.47 for an example field shape allowed with MLCs. Although the light field may appear to have very jagged edges, the radiation field edges are much smoother inside the patient because of the effect of scatter. The use of MLCs has largely replaced the lengthy and laborious task of creating patient-specific custom blocks.

The MLC banks can either be positioned externally in addition to the standard collimation system, or they can replace one of the secondary collimator jaws. However backup collimation will normally be needed with MLCs due to the significant interleaf leakage that occurs due to the necessary mechanical clearance between each leaf. This is minimised by using a tongue and groove design, which is illustrated in Figure M.48. The end of each leaf is tapered so that their edges are focused towards the target in order to minimise the penumbra irrespective of the angle of incidence. The positional systems vary between manufacturers, but include the use of potentiometers or an optical system with reflectors on each leaf.

**MLCs and IMRT:** MLCs are used in the delivery of intensity-modulated radiotherapy (IMRT) enabling the fluence of the beam to be modulated over time and position. There are two methods, the multiple static field (MSF) technique, and the dynamic MLC technique. The MSF technique uses the MLCs to create a sequence of 2D-shaped fields, only moving the MLC when the beam is off. The dynamic MLC method uses continuously moving leaves whilst the radiation is on. Varying the velocity of each leaf pair can create any required 2D profile.

**MLC QA:** QA considerations for MLCs include the following:

- Leaf position accuracy
- Collimator leakage (tongue and groove leakage)
- Leaf speed control
- Interlocks

**Abbreviations:** IMRT = Intensity-modulated radiotherapy, MLC = Multileaf collimator, MSF = Multiple static field and PTV = Planning target volume.

**Related Article:** Intensity-modulated radiotherapy (IMRT)

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Multimodality systems
*(Nuclear Medicine)*

A multimodality system refers to an imaging system where two or more scanners with a single patient table share the same system of reference. Examples of dual-modality
systems are PET-CT and SPECT-CT-scanners. With such a system it is possible to sequentially acquire a functional and morphological image. CT scans can also be used for attenuation correction.

When combining images from stand-alone scanners the patient has to move between the two scanners. It is not likely that the patient can assume an identical position in the second scan and also the organ orientation differs slightly between the two scans. These sources of error are avoided when using a multimodality system and can therefore lead to higher diagnostic accuracy. For example, using a PET/CT system for tumour localisation and diagnosis has improved the diagnostic accuracy by 48%–60%. It has affected the clinical management decisions in 12%–27% compared to stand-alone PET and CT and has lead to a modification in up to 63% of patient external radiation treatment plans.


Multi-pinhole collimator
(Nuclear Medicine) This collimator includes several pinhole apertures instead of only a single hole. A pinhole collimator is primarily used because it provides images with high spatial resolution. The pinhole collimator efficiency, however, depends on the size of the apertures and the source to collimator distance. At distances smaller than ~60 mm the pinhole efficiency (2 mm apparatus opening) is often better or equal to the efficiency of a parallel-hole collimator. In vivo distributed sources are often located further away from the collimator than 60 mm, thus giving a low efficiency for clinical studies. Low efficiency means long acquisition times and low signal to noise ratio.

A way to increase the efficiency is to use multiple pinhole collimators since multiple pinholes allow simultaneous acquisition of different projections. The increase in efficiency then is a factor of the number of pinhole apertures but also of the crystal size and thickness. When the number of pinholes increases there is also an increased probability to acquire projections that overlap which may hamper an accurate image reconstruction. A proper collimator design therefore depends on a good optimisation between these two factors.

Related Article: Collimator design

Multiplanar reconstruction (MPR)
(Magnetic Resonance) A magnetic field can be established in any direction in space by applying currents of the appropriate amplitudes to two or three of the gradient coils simultaneously. Thus it is possible to collect images in any plane, as long as the net slice selection, frequency encoding and phase encoding gradients produced in this way remain mutually orthogonal.

Nevertheless, it is sometimes useful to be able to reformat image data acquired in a particular plane for viewing in a different plane. This process of displaying images in different planes is known as multiplanar reconstruction.

Where, as is often the case, two-dimensional images have been collected using a slice thickness considerably greater than the in-plane resolution, reconstruction in a different plane can result in stair-step artefacts. This is usually less apparent with three-dimensional imaging, where there is in principle no preferred plane and voxels are more likely to be isotropic.

Related Article: Oblique imaging

Multiple beams
(Radiotherapy) Multiple beams are used in conformal radiotherapy, in order to conform the dose distribution as closely as possible to the target volume.

Related Article: Conformal radiotherapy

Multiple coulomb scattering
(Radiation Protection) When charged particles in a beam of ionising radiation are incident on an absorbing medium they will undergo not just one, but a series of scattering interactions before all the energy in the incident particles has finally been absorbed. These scattering events are due to the interaction between the charge of the particles, and the electrostatic (Coulombic) forces present in the nuclei of the atoms of the medium. These interactions are also known as Rutherford scattering.

Multiple isocentre treatment
(Radiotherapy) In most radiotherapy treatments, the centre of the tumour (target) is positioned at the isocentre. However, in some cases more complex target volumes or multiple targets located close together necessitate consideration of overlapping dose distributions across the entire volume using more than one isocentre location.

This multiple isocentre technique is used most often with the gamma knife treatment system. This uses a number of cobalt sources arranged in a hemisphere which are focused using a collimator helmet onto a central target and will produce small, spherical, high dose volumes. By combining these small radiation spheres, positioned at multiple isocentres, it is possible to conform to the shape of larger and non-spherical tumours.

Multiple reflection
(Ultrasound) See Reverberation

Multiple scan average dose (MSAD)
(Diagnostic Radiology) The multiple scan average dose (MSAD) is used in computed tomography (CT). It is a measure of average absorbed dose (mGy) to the irradiated area from a series of slices. It is a consequence of the fact that the dose profile of each CT scan is not perfect (square), but is bell-shaped (for details see the articles on Dose profile). The need of MSAD manifests itself when a sequence of adjacent CT scans is made. Cases where the couch increment is equal or less than the nominal slice width the bell-shaped edges of the dose profiles of each CT scan overlap, and result in the average (summed) dose being higher than the dose in a single CT scan. (Figure M.49). Obviously the higher the spread of the bell shape of the dose profile, the higher the MSAD.

For cases where the couch increment is greater than the slice width, the MSAD is reduced and may be lower than the dose from a single scan (Figure M.50).

The formula used to calculate MSAD is

$$MSAD = \frac{1}{I} \int D(z)dz$$

where

$D(z)$ is the dose profile along the z-axis (scanner axis of rotation)
$I$ is the scan increment between the adjacent slices

The MSAD for a scan length of 100 mm is equal to CTDI<sub>100</sub> (the commonly used metric in CT dosimetry) when the couch increment is equal to the nominal slice width, or, in helical scanning, when the pitch
value is equal to unity. For other cases, the CTDI must be corrected for couch increment or pitch to obtain the MSAD (see the following):

\[
\text{in axial scanning: } MSAD = \text{CTDI} \left[ \frac{T}{I} \right] \\
\text{in helical scanning: } MSAD = \text{CTDI} \left[ \frac{1}{\text{Pitch}_x} \right]
\]

**Related Articles:** Dose profile, CTDI

**Multiple scattering** *(Radiation Protection)* A proportion of the photons or charged particles in a beam of ionising radiation incident on an absorbing medium will undergo not just one, but a series of scattering interactions before all the energy in the incident radiation has finally been absorbed. Such multiple scattering events can spread the energy absorption some distance away from the primary radiation beam, having an implication for radiotherapy treatment planning, etc., where the goal is to deliver a prescribed radiation dose to the target volume (the tumour) whilst minimising the radiation dose to surrounding healthy tissues.

**Related Articles:** Elastic scatter, Inelastic scatter

**Multislice** *(Magnetic Resonance)* Acquisition of a single-slice image through a body structure of interest is rarely sufficient for diagnostic purposes. Instead, it is usual to acquire either a stack of slices or a volume of data.

Because of the nature of the data acquisition process in MRI, acquisition of \( N \) slices does not normally entail sequential repetition of the acquisition process \( N \) times. The need to allow a repetition time \( (T_R) \) for recovery of longitudinal magnetisation between consecutive excitations of a given slice fortuitously provides a ‘dead time’ during which other slices may be interrogated. Depending on the \( T_R \) used (i.e. the desired degree of \( T_1 \) weighting) it may be possible to image a considerable number of slices without an increase in the overall imaging time.

A significant complication arises because the profile of an excited slice in MRI never corresponds exactly to the desired ‘top hat’ function, and there is inevitably some excitation of material lying outside the ideal slice. As a result, if consecutively excited slices (e.g. A and B in Figure M.51) are contiguous, then material lying close to the boundary between the slices will be repeatedly excited at very short intervals. This results in heavy \( T_1 \) weighting of this material and severe distortion of the slice profile.

There are two possible solutions to this problem, other than improving slice profiles by computational modelling. A common approach is to leave a small gap between slices, often 10% of the slice thickness, to allow for profile imperfections. This has the disadvantage of possibly missing clinically significant information lying in the slice gap. Alternatively, two interleaved acquisitions may be performed, acquiring odd and even numbered slices respectively. This restores slice contiguity, but requires a doubling of the overall imaging time.

![FIGURE M.49](See colour insert.) MSAD for case where couch increment equals slice width.

![FIGURE M.50](See colour insert.) MSAD for case where couch increment is greater than slice width.

![FIGURE M.51](Acquisition of signals from multiple slices within a single \( T_R \) interval.)
Multislice imaging is more complicated when sequences are used that acquire multiple lines of $k$-space from the same slice following a single excitation (e.g. fast spin echo). In such cases the echo train length and the number of slices imaged in a single acquisition may need to be traded off to ensure optimal use of the $T_2$ interval.

**Related Articles:** Fast spin echo (FSE), Interleaving, Slice selection

### Multislice (Nuclear Medicine)

Multislice is a camera configuration with several detector rings (or slices). Each detector ring often consists of several individual camera elements, each coupled to a signal multiplier, for example a PM tube.

Since most conventional scintillation cameras have individual camera heads instead of a set of rings, multislice cameras in nuclear medicine are often referred to as PET cameras. A multislice PET system can be operated in either a two- or three-dimensional mode (2D and 3D mode respectively). 2D mode involves using a separating high attenuating material between each ring, that is septa that attenuates radiation with oblique angle of incidence. When operated in a 2D mode, the PET camera will only accept annihilation photons with a near-perpendicular direction relative to the detector ring surface, that is both photons must be registered in the same ring (see Figure M.52). Not all annihilation photons have a near-perpendicular incident angle and are therefore attenuated by the septa or simply discriminated by the coincidence processor. In a 3D system there are no septa and coincidences can be registered between all rings; hence a higher sensitivity. For further details about PET acquisition, see Related Article.

**Related Article:** Data acquisition PET


### Multislice CT scanner (Diagnostic Radiology)

For many years the detection system on CT scanners consisted of a single arc of detectors (Figure M.53a) and so acquired one slice of data in a gantry rotation. In 1991 a scanner with two parallel banks of detectors, capable of acquiring two slices simultaneously was launched. This paved the way for modern multislice systems, and in 1998 the first four-slice scanner with multiple parallel banks of detectors (Figure M.53b) was announced. This was followed by 8-slice scanners in 2001, 16-slice scanners in 2002, 64-slice scanners in 2004, and in 2007, a scanner capable of acquiring 320 slices per rotation (Figure M.54). The increased length per rotation covered by these scanners, coupled with faster gantry rotation speeds, allows the scanned volume to be covered very quickly, thereby largely removing motion artefacts. On these systems the scan length is no longer limited by factors such as the patient breath-hold time.

On single-slice scanners the acquired slice width can be varied between 1 and 10mm by varying the $z$-axis x-ray beam collimation. On multislice systems the width of the acquired slice, sometimes referred to as the ‘data acquisition width’, is governed by the $z$-axis detector dimension. Data from multiple adjacent channels can

**FIGURE M.52** Two different acquisition modes, 2D (upper) and 3D (lower). When operating in a 3D mode the scanner has higher sensitivity than when running in a 2D mode.

**FIGURE M.53** (a) Single-slice scanner and (b) multislice scanner (not to scale). (Graphs courtesy of ImPACT, UK, www.impactscan.org)
be combined and this results in an increase in the data acquisition width. An example of a four-slice scanner is shown in Figure M.55. The scanner has 16 parallel, 1.25 mm detector banks, but can only acquire a maximum of four data channels simultaneously. For z-axis beam collimations in excess of 5 mm, the signals from adjacent detector banks are combined, resulting in an increased data acquisition width. Scanners capable of acquiring a high number of data channels per rotation, such as 64-slice scanners, do not usually require this combining of signals, and can therefore always acquire data with narrow acquisition widths, even for the widest beam collimations.

In addition to the increased coverage per gantry rotation, the second major advantage of multislice scanners is that they can acquire a wide coverage of data, at widths as narrow as 0.5 mm. This capability means that narrow slice acquisition can be used for most examination types as it no longer results in unacceptably long scan times or excessive tube loading. Narrow slices result in improved z-axis spatial resolution and provide isotropic resolution images, which provide images of equal quality when reconstructed in any plane (Figure M.56a). They also enable high quality 3D reconstructions (Figure M.56b).

Multislice CT has extended the range of clinical examinations possible with CT scanners, which has lead to concerns about the radiation dose burden from these scanners, so CT examinations must always be justified. And although scanning long length with high tube currents is possible, effort must be made towards optimisation of scan parameters.

**Related Articles:** Computed tomography, Helical scanning, Slice thickness, Image artefact

**Mu-metal**

*(Diagnostic Radiology)* Mu-metal is a special alloy used for magnetic shielding. It is a nickel–iron alloy (75% nickel, 15% iron, plus copper and molybdenum) with very high magnetic permeability. One specific use of this alloy is the envelope of Image Intensifiers in fluoroscopy, where the precise internal electromagnetic filed requires good shielding from external electromagnetic influences.

**Muscle**

*(General)* Muscle is around 70% water, with the remaining 30% composed mostly of protein in the form of actin and myosin. Muscles can be classified into three groups: skeletal, smooth or cardiac. Skeletal muscle is attached to bone and is responsible for skeletal movements and posture. Smooth muscle exists within organ walls and controls involuntary movements such as peristalsis in the oesophagus. Cardiac muscle is found in the heart. Skeletal and cardiac muscles are striated: the muscle fibres are divided into sections or sarcomeres, where thin filaments formed by actin, troponin and tropomyosin overlap to regularly varying degrees with thick filaments formed of myosin. During muscle contraction, the thick and thin filaments slide past each other to shorten the muscle fibre.

**Properties of Muscle-Based Tissues for Medical Imaging:** Muscle can be distinguished on x-ray images, although...
contrast with other soft tissues is generally poor. Muscle has a CT number of between +10 and +40 Hounsfield Units. The distinction of muscle from other soft tissues is much better in magnetic resonance imaging. Muscle has $T_1$ and $T_2$ relaxation times of around 1075 and 33 ms respectively, at a magnetic field strength of 1.5T. Muscle has a mass attenuation coefficient of 0.0227 m$^2$/kg$^1$ for 50 keV photons.

Related Article: Tissue contrast

Myocardial perfusion imaging

(Magnetic Resonance) Perfusion is an assessment of the capillary blood flow through a tissue. Myocardial perfusion scans show the blood supply to the cardiac muscle (the myocardium), and can visualise regional defects in the perfusion, caused by a stenosis in the supplying coronary vessel, or an infarction. Such perfusion defects can demonstrate delayed perfusion, less perfusion, or a combination of these, in comparison to the well-perfused myocardium. The perfusion is often evaluated in both rest and stress, where the stress can be either mechanical or pharmaceutical. In the clinical setting, assessment of myocardial perfusion defects is so far qualitative.

Myocardial perfusion is measured by single photon emission computed tomography (SPECT) and positron emission tomography (PET) with good results, but these modalities expose the patient to ionising radiation. In MRI, myocardial perfusion is assessed by injecting a small Gd-bolus and imaging as the bolus passes the heart, the so-called first-pass perfusion. The imaging sequence uses the difference in $T_1$ introduced by the passing Gd-bolus, and $T_1$-sensitivity is achieved by using saturation recovery. To be able to resolve the fast course of events, a read-out with adequate time resolution is necessary. Fast gradient echo (GRE), GRE-EPI or balanced GRE readouts are used. 2–4 slices are acquired, and all slices should be acquired in each heartbeat. The imaging sequence takes about a minute, and for good image quality, the patient should maintain breath-hold for as long as possible, and at least during the bolus passage.

Related Articles: Dynamic susceptibility contrast MRI, Perfusion imaging

Myo-inositol

(Magnetic Resonance) Myo-inositol is a chemical compound that features in in vivo proton ($^1$H) NMR spectra of the brain (Figure M.57).

![Figure M.56](See colour insert.) (a) Sagittal multi-planar reconstruction of ankle bones. (Courtesy of Siemens Medical Systems.) (b) 3D volume rendered image of kidneys and associated vasculature. (Courtesy of GE Healthcare.)

![Figure M.57](Molecular structure of myo-inositol.)

![Figure M.58](1H NMR spectrum of the human brain showing main myo-Ino resonance.)

The main myo-inositol resonance occurs at 3.54 ppm, and at magnetic field strengths higher than those used in clinical MRI this is resolved into a complex multiplet. There are also smaller resonances at 3.28, 3.60 and 4.05 ppm.

Myo-inositol has a number of biological functions, mostly related to cell signalling. High levels of myo-inositol are found in neonatal brain spectra, and in adulthood levels are elevated in Alzheimer’s disease and reduced in hepatic encephalopathy. It is widely regarded a marker of astrocytes, as changes in concentration are also seen in certain types of tumour (Figure M.58).
N/2 artefact

(Magnetic Resonance) The N/2 artefact is another term for the Nyquist ghosting artefact found only for the EPI pulse sequence.

**Abbreviation:** EPI = Echo planar imaging.

**Related Articles:** Ghost artefact, Nyquist ghosting

N-acetylaspartate (NAA)

(Magnetic Resonance) N-acetylaspartate (NAA) is a chemical compound that features in proton (H) NMR spectra of the brain. The methyl proton resonance from NAA at 2.01 ppm is the most prominent peak in a water-suppressed spectrum, and a smaller multiplet resonance at 2.6 ppm due to $\text{CH}_2$ protons may also be visible (Figure N.1).

In the adult human brain, NAA is found only in neuronal bodies. Its role is poorly understood, but most likely it has an osmoregulatory function. Despite this uncertainty, NAA has an important role in MR spectroscopy, since it is regarded as a neuronal marker, and its depletion is thought to be indicative of a loss of functional neurons. For example, depletion of the NAA peak at 2.01 ppm is found in degenerative disorders such as Alzheimer’s disease, where this change may occur before cerebral atrophy becomes visible on MRI. In temporal lobe epilepsy, reduced NAA can help to lateralise the disease in the absence of MRI findings – although there is also a contralateral NAA depletion that may resolve following surgery to the ipsilateral focus, indicating that reduction in NAA is associated with poor neuronal function, rather than simply a loss of neurons (Figure N.2).

**Related Article:** Magnetic resonance spectroscopy


NaI(Tl) detector crystal

(Nuclear Medicine) The crystalline sodium iodine with a thallium impurity is the ‘workhorse’ crystal used in emission imaging. It was introduced in 1948 by Robert Hofstadter and it demonstrated an exceptionally large scintillation light output compared to the organic scintillators previously used. The crystal has very few obvious drawbacks (only in certain applications). NaI(Tl) plays a central role (or has) in nearly all clinical fields of nuclear medicine and is considered as the standard scintillation material.

The most favourable property of the NaI(Tl) scintillator is the high light yield, namely 38,000 scintillation photons per MeV. As most inorganic scintillators, NaI(Tl) shows non-linearity in light yield, especially at low energies. NaI(Tl) is also hygroscopic and sensitive to mechanical and thermal stress so the crystal must be sealed properly. The dominant decay time is 230 ns but there is also a phosphorescence component with a decay time of about 0.15 s that is as large as 9% of the total light yield, that is 9% of all de-excitations are delayed because the electrons get caught in energy state were further de-excitation is forbidden. The electron must first be excited to an adjacent upper energy state before further de-excitation. The delayed component causes a number of low signal amplitude events which is generally (at low and medium count rates) no problem since they are discriminated by the detection electronics due to low signal amplitude. There are applications in which single-electron sensitivity is needed and the result is degraded by the phosphorescence pulses. At high count rates the phosphorescence tends to build up when a large number of phosphorescence pulses overlap.

**Related Articles:** Inorganic scintillators, Scintillators


Narrow beam geometry

(Nuclear Medicine) Narrow beam geometry refers to situations in which a radiation beam profile is typically smaller than a few cm². The opposite situation is referred to as broad beam geometry. Consider two different beam geometries falling on a perpendicular surface. The dose along the central line in the two cases will differ since in the broad beam situation photons outside the central line can scatter inwards towards it (sometimes referred to as ‘in-scatter’) and contribute to the dose. The dose along a central line is therefore higher in the broad beam geometry compared to a narrow beam of equal intensity per unit area.
National Council on Radiation Protection and Measurements (NCRP)

(Radiation Protection) The National Council on Radiation Protection and Measurements (NCRP) started its activity in 1929 as ‘The Advisory Committee on X-ray and Radium Protection’; originally it was established to represent all of the national radiological organisations in the United States. The idea was to create in the United States the national analogue of the International X-Ray and Radium Committee, created in 1928 (see ICRP). Due to the extreme increase in the use of ionising radiation, the initial informal approach was not considered sufficient and, in 1964, the committee was reorganised and formalized by the US Congress.

The NCRP is an US organisation with the scope to formulate and disseminate information, guidance and recommendations on radiation protection and measurements. The council closely follows the publications of the International Commission on Radiological Protection (ICRP) with the focus on data and material which can make a contribution of public interest.

The council has also the mission to facilitate and stimulate cooperation among organisations dealing with radiation protection. One of the main activities is to publish reports on the various topics.

Hyperlink: NCRP. http://www.ncrp.org

National Radiation Authority (NRA)

(Radiation Protection) See Regulatory authority

National Radiological Protection Board (NRPB)

(General) The National Radiological Protection Board (NRPB) merged in 2005 with the UK Health Protection Agency (HPA) and is its Radiation Protection Division. The Division consists of Headquarter at Chilton in Oxfordshire; Occupational Services Department in Leeds and Radiation and Environmental Monitoring Scotland at Glasgow. Jointly with the HPA division for Chemical Hazard and Poisons it forms the Agency Centre for Radiation, Chemical and Environmental Hazards.

The Radiation Protection Division has an Advisory Group on Non-ionising Radiation (AGNIR) and an Advisory Group on Ionising Radiation (AGIR). They are responsible for the Agency’s work on non-ionising and ionising radiations, including research to advance knowledge about protection from the risk of these radiations. It also provides laboratory services, organises training courses and provides expert services.

The HPA (UK Health Protection Agency) Radiation Protection Division (formerly NRPB) has always had a leading position in the field of medical applications and has a significant advisory role in the United Kingdom.

Hyperlink: http://www.hpa.org.uk

Navigator echo

(Magnetic Resonance) Navigator echoes are used in MRI to correct for or prevent motion artefacts occurring due to object displacement or motion-induced phase shifts. Commonly it is used for prospective acquisition correction (PACE) of, for example abdominal imaging or in multi-shot diffusion weighted imaging where macroscopic motion severely degrades the image quality.

The navigator echo is a non-phase encoded echo, sampled after the same excitation pulse as the echo that is to be used for imaging, but can be collected before or after the actual image echo. Phase corrections utilising so-called navigator echoes are based on the assumption that if two echoes, an image echo and a navigator echo, recalled after the same RF excitation pulse, are collected within a short period of time, they will be influenced by the same amount of macroscopic motion and will thus exhibit the same motion-induced phase shift. In the case when no motion is present, the phase of the MR signal for each line in k-space should be the same, but when an echo is subjected to motion this will affect the phase of that echo. By determining a reference phase for one line (one echo) the following echoes can be phase shifted based on the phase for the corresponding navigator echo. Correction of phase errors due to translational motion is corrected directly in the k-space domain whereas phase correction of the MRI signal occurring due to rotational motion should be performed on projection data.

For prospective acquisition correction the signal from a one-dimensional (1D) ‘rod of tissue’ is collected for the same excitation period as the image data. Again we are assuming that the navigator echo and the image echo are affected by the same motion pattern. The size of the tissue column is then used to determine whether the body can be considered to be in the same position as in previous excitations and data can be either accepted or discarded.

Related Article: k-space


Negative contrast media

(Diagnostic Radiology) The negative contrast media used in x-ray imaging has lower absorption of x-rays (such as air, carbon dioxide, etc.). It allows easy penetration of the x-rays and presents the object filled with this contrast as a darker area on bright background (this refers to an image on x-ray film). Negative contrast media can be used in conjunction with positive-contrast media (high absorbent media, such as iodine, barium meal, etc.) to produce double contrast. Typical use of such contrast is in gastro-intestinal examinations.

Digital x-ray systems allow contrast inversion as part of image processing, which should not be mistaken with the use of negative contrast media. Often the visual perception of darker area on bright background is better, than the reverse contrast. This human phenomenon is used, for example in the so-called positive-contrast mammograms.

Negative contrast media

(Magnetic Resonance) This term refers to contrast agents that make particular tissues more conspicuous by decreasing the signal from them.

In MRI, image contrast results from interplay between the NMR properties of hydrogen nuclei (protons) in tissue and pulse sequence parameters. Negative contrast is achieved by using a $T_2$- or $T_2^*$-shortening agent, together with a $T_1$- or $T_2$-weighted pulse sequence. The shortened relaxation times of protons in regions receiving a high concentration of the agent result in faster signal decay and hence decreased signal.

Gadolinium chelates in sufficient concentration have a $T_2$-shortening effect, particularly those based on dysprosium. However, negative contrast agents are more commonly based on small superparamagnetic particles of iron oxide. These affect relaxation primarily through outer sphere effects, resulting in dramatic reductions in $T_2$. As the particles become larger, it is a matter of semantics as to whether the effect is regarded as a shortening of the relaxation time or irreversible dephasing due an increase in local field inhomogeneity – that is, a susceptibility effect.
Most contrast agents are administered intravenously. Simple agents such as carbon dioxide, barium sulphate and perfluorochemicals may be administered orally for suppression of bowel signal which may otherwise obscure structures of interest and lead to motion artefacts. However, these agents may stimulate peristalsis which is counter-productive. There are commercially available bowel suppression agents, again usually based on superparamagnetic particles.

**Related Articles:** Gadolinium chelate, Paramagnetic contrast agents, Ultrasmall particles of iron oxide (USPIO), Superparamagnetic particles, Superparamagnetic iron oxide, Positive-contrast media

**Negative-ion cyclotron**

 *(Nuclear Medicine)* A device in which negatively charged particles, for example H⁻ (hydrogen atom with two electrons) are accelerated in circular paths to several MeV in a magnetic field. In the case of H⁻, the negatively charged high energy particle (in the MeV range) loses two electrons when passing through a thin foil of carbon prior to extraction. Hence the total particle charge is converted from a negatively charged hydrogen atom to a positive proton P⁺. This polarity change is used to extract the proton beam since P⁺ have a curvature opposite to the H⁻. The extraction is illustrated in Figure N.3 where protons are extracted at two different locations from different parts of the H⁻ beam. After the extraction, the beam is directed onto a target.

**Negative pions**

 *(General)* Negative pions are one of the three types of subatomic particles known as pions, which is an abbreviated form for the pi meson. The symbol for the negative pion is the \( \pi^- \) and for the positive pion \( \pi^+ \). Pions have zero spin and are composed of quarks. The negative pion has a mass of 139.6 MeV/c², a mean life of \( 2.6 \times 10^{-8} \) s and negative charge. It decays into a muon and neutrino.

The use of the negative pions has been explored in radiotherapy due to the depth-dose pattern of a negative pion beam and other characteristics.

**Related Articles:** Auger particles, Beta particles, Compton scattering, Internal conversion electrons, Photoelectric effect

**NEMA**

 *(Ultrasound)* NEMA, the National Association of Electrical and Medical Imaging Equipment Manufacturers, represents the industry’s interests in ultrasound and other imaging modalities. NEMA produces guidance and recommendations for the safe and effective application of medical ultrasound. In association with the AIUM, NEMA issued the Output Display Standards (ODS) safety displays and recommendations for diagnostic ultrasound.

**NEQ (noise equivalent quanta)**

 *(Diagnostic Radiology)* See Noise equivalent quanta

**Nerve stimulation**

 *(Magnetic Resonance)* See Peripheral nerve stimulation (PNS)

**Net magnetisation**

 *(Magnetic Resonance)* When a water sample is introduced into a magnetic field \( B_0 \), the magnetic moment (here denoted spin) of each hydrogen nucleus in the sample experiences two effects derived from quantum mechanics: The spins align with the main magnetic field direction (the \( z \) or longitudinal direction) in one out of two possible ways, either aligned with the +z direction (spin-up, \( N_+ \)) or along the −z direction (spin-down, \( N_- \)). The alignment is in either case made with an angle relative to the z-axis (see Figure N.4). Simultaneously, the spins start to rotate around the z-axis (precession) with the Larmor frequency \( 42.6 \text{MHz/T} \). It can be shown that a small excess (on the order of parts per million, ppm) of the spins are aligned in the +z direction. The excess amount depends upon, for example temperature and \( B_0 \). As an example, the polarisation \( P \) defined as

\[
P = \text{ABS} \left( \frac{N_+ - N_-}{N_+ + N_-} \right)
\]

is approximately \( 5 \times 10^{-6} \) at 1.5 T and at body temperature.

In an MRI experiment, a vast number of spins contribute to this process in each imaging element (voxel) and since the rotations of the individual spins are not synchronised (not in phase), a net magnetisation vector \( (M_0) \) is built up in the z-direction – see Figure N.4.

As mentioned, \( P \) and hence also the magnitude of \( M_0 \) depends upon the main magnetic field, and when \( B_0 \) increases \( M_0 \) increases. In MRI, \( M_0 \) is used to create detectable signal by ‘flipping’ the magnetisation into the transversal plane under the influence of an external RF field and hence, an increase in \( B_0 \) creates an increase in signal-to-noise ratio (SNR). The relation between \( B_0 \) and SNR is generally recognised to be approximately linear in the range of \( B_0 \) values applicable to clinical MRI.

**FIGURE N.3** Extraction of protons is accomplished by stripping negatively charged hydrogen atoms of their two electrons. Protons with opposite charge will bend outwards and can then be directed onto a target.

**FIGURE N.4** Net magnetisation.
Neutrino

There are three types of neutrino – electron, muon and tau. Their symbols are \( \nu_e \), \( \nu_\mu \) and \( \nu_\tau \). They have no charge and a very small mass, which has not been measured. There are also anti-neutrinos (symbol \( \bar{\nu} \)) for the electron anti-neutrino.

Electron anti-neutrinos are emitted from the nucleus when protons are changed into neutrons in a nuclear transformation. An example is the beta decay of the radionuclide phosphorous-32 into sulphur-32, by the transformation of a neutron into a proton with the emission of a beta particle (electron) and an anti-neutrino: \( \nu_{e}^{0} \rightarrow \nu_{e}^{+} + \beta + \nu_{e}^{0} \).

Often the neutrino is omitted from decay equations.

In positron decay a proton in the nucleus is converted into a neutron with the emission of a positron (positive electron or beta particle) and a neutrino.

The existence of the neutrino was first proposed by Wolfgang Pauli in 1930 to explain the conservation of energy in beta decay and neutrinos were only experimentally detected in 1956.

**Related Articles:** Beta decay, Positron

Neutron capture therapy

**(Radiotherapy)** Neutrons, having no charge, are highly penetrating, but lose their energy by collisions with hydrogen nuclei in tissue. Fast neutrons (5–30 MeV) from accelerators have been used in external beam neutron therapy for cancer, and found to have an advantage in slow growing cancers such as salivary glands. However, the hydrogen scattering leads to poorly collimated beams, and problems for normal tissue tolerance. Neutrons also have complications arising from late radiation effects.

Neutrons in biology

The neutron is a subatomic particle having no charge and a mass that is about 1 atomic mass unit (amu), or 1.67 × 10⁻²⁷ kilograms. In 1932, James Chadwick discovered the neutron, which was previously incorrectly believed to be a composite of protons and electrons. Neutrons are constituents of the atomic nucleus and can exist in an atomic nucleus or in a free state. In a biological system, the neutron is found mainly in the nuclei of hydrogen atoms, usually in the form of the tritium isotope, or in the nuclei of heavier elements such as carbon and oxygen.

Neutrons are identical to protons except that they have zero electric charge, and neutrons can therefore pass through the atomic nuclei that make up the atoms that make up living organisms. Neutrons are therefore important in biological processes that involve the movement of the elements comprising those atoms, although few biological systems are specifically designed to take advantage of the properties of neutrons. However, the discovery of the neutron has led to important developments in the field of biology.

**Neutron capture therapy** (Radiotherapy) Neutrons, having no charge, are highly penetrating, but lose their energy by collisions with hydrogen nuclei in tissue. Fast neutrons (5–30 MeV) from accelerators have been used in external beam neutron therapy for cancer, and found to have an advantage in slow growing cancers such as salivary glands. However, the hydrogen scattering leads to poorly collimated beams, and problems for normal tissue tolerance. Neutrons also have complications arising from late radiation effects.

**Neutron activation** (Radiotherapy) Neutron activation treatments involving energies higher than the threshold needed to liberate a neutron may result in neutron activation, particularly for low atomic number materials placed in the beam. Examples of reactions leading to neutron activation include the \( (\gamma,n) \) reaction for therapeutic x-ray beams and \( (p,n) \) reaction for proton beams. The threshold energy for the \( (\gamma,n) \) reaction is generally at the higher energies used for external beam treatment, for example the threshold for \( O^{16} (\gamma,n) O^{15} \) is 15.7 MeV (here x-rays and \( \gamma \)-rays are treated synonymously). For hadron beams such as protons, a range of activation products may be created in addition to neutrons, including other changed particles, but these do not have the path lengths of neutrons.

**Related Articles:** External beam radiotherapy, Hadron therapy, Proton therapy

**Neutron capture cross section** (General) Neutron capture is one of the two processes that a neutron can interact with the nucleus of an atom – the other is neutron scattering. Neutron capture may lead to either absorption or fission.

The cross section is a measure of the probability of an interaction and can be visualised as the cross-sectional area presented to an incident neutron. The neutron capture cross section is the number of neutrons captured per unit volume per second for a unit incident flux of neutrons and unit nuclear density.

If the symbol \( \delta \) (Greek delta) is used to denote the cross section, then \( \delta = \pi R^2 \) cm², where \( R \) is the radius of the nucleus. A particle which passes through a thin sheet of material of area \( A \) containing \( N \) nuclei which do not overlap will have a probability of \( N \delta / A \) of colliding with a nucleus. The number of nuclei/cm², called the surface density, given by \( N / A \) is equal to \( N_t \), where \( N \) is the number of nuclei/cm³ and \( t \) is the thickness of the sheet. The collision rate of an incident beam of \( n \) neutrons/cm², which have a velocity of \( v \), is \( n \delta / A = n \delta / V \). The cross section is the number of collisions/cm²/s divided by \( n / V \), \( n \) is the particle flux – the number of particles crossing 1 cm² of area every second.

The unit barn (b) can be used to express the cross section and 1 barn = 10⁻²⁴ cm².

Each separate capture process, denoted by \( i \), will have an individual cross section which can be summed to a macroscopic cross section: \( \Sigma = N \delta_i \).

**Related Articles:** Nucleus, Neutron activation

**Network architecture** (General) Network architecture refers to the design of a computer network, either a local area network or a larger installation. This can include hardware or software and protocols such as the commonly known SMTP, POP3, DHCP, HTTP, TCP/IP, FTP and Telnet.

The hardware design for a LAN would include the topology; the pattern of linking each computer or network node. Such designs include closed loop or ring cable connections, single line or bus connections, star, mesh or tree connections or fully interconnecting designs (Figure N.5).

**Neuroreceptor targeting** (Nuclear Medicine) This term refers to radiopharmaceuticals that target receptors in the neurological system. Neurilgands can be labelled with suitable radioisotopes and used to study different neurological systems, for example the dopaminergic system. Such radiopharmaceuticals can be used to study neurodegenerative diseases, such as Parkinson’s disease.

**Related Articles:** Tracer kinetic modelling, Receptor targeting, Antigen targeting, DNA targeting, Glycolysis targeting, Apoptosis targeting, Hypoxia targeting


**Neutron conductor** (General) The conductor in a two-wire (single phase) AC electrical system intended for current flow back to the source. It is usually connected to the earth (ground) at the source, that is at the low voltage site of the mains power transformer. In an ideal case, neutral and earth would be at the same potential. In reality, the neutral is at a higher potential, because of the voltage drop on the neutral conductor due to the return current flow.

In some cases, when the local ground of the electric supply system is connected to the neutral, the voltage at conductive enclosures of equipment or appliances may become dangerous. Connection of earth to neutral conductor is not allowed in medical equipment.

**Related Article:** Earthing (ground)

**Neutrino** (General) A neutrino is an elementary particle which has no charge and a very small mass, which has not been measured. There are three types of neutrino – electron, muon and tau. Their symbols are respectively \( \nu_e \), \( \nu_\mu \) and \( \nu_\tau \). There are also anti-neutrinos (symbol \( \bar{\nu} \)) for the electron anti-neutrino.

Electron anti-neutrinos are emitted from the nucleus when protons are changed into neutrons in a nuclear transformation. An example is the beta decay of the radionuclide phosphorous-32 into sulphur-32, by the transformation of a neutron into a proton with the emission of a beta particle (electron) and an anti-neutrino: \( \nu_{e}^{0} \rightarrow \nu_{e}^{+} + \beta + \nu_{e}^{0} \).

Often the neutrino is omitted from decay equations.

In positron decay a proton in the nucleus is converted into a neutron with the emission of a positron (positive electron or beta particle) and a neutrino.

The existence of the neutrino was first proposed by Wolfgang Pauli in 1930 to explain the conservation of energy in beta decay and neutrinos were only experimentally detected in 1956.

**Related Articles:** Beta decay, Positron
Neutron therapy

Fast neutrons from reactors or accelerators can be moderated and thermalised by collisions with hydrogen nuclei. Some isotopes have very large capture cross sections for thermal neutrons (0.025 eV). For example, Gd-157 (15.68% natural Gadolinium) has an exceptionally high thermal neutron capture cross section (240,000 barn), and can cause double strand breaks of DNA from high LET auger electrons after thermal neutron capture. Boron-10 (19.78% natural Boron) has a high thermal neutron cross section of 3840 barn and high LET products that are highly toxic to targeted cells. Therefore, if boron-10 atoms are secreted into tumours (∼30 ppm), then incident thermal neutrons will be captured by B-10, followed by the emission of very short range (around 10 μm, the order of one cell diameter), high LET and highly toxic radiation:

\[ ^{10}\text{B} + n \rightarrow ^{4}\text{He} + ^{7}\text{Li} + 2.31 \text{MeV} \ (93.9\% \text{RBE}) \]

Thermal neutron capture therapy utilising this process is called boron neutron capture therapy (BNCT). Many different boron compounds have been tested. The compound of choice is the amino acid analogue 2-p-boronophenylalanine, solubilised at neutral pH by complexation with fructose (BPA-F).

While skin melanomas have been treated by thermal neutron capture therapy, and brain tumours treated by intracavitary thermal neutrons, deep seated tumours require an epithermal neutron beam (∼5 keV) to give sufficient penetration for a thermal flux of ∼10¹³ neutrons cm⁻² s⁻¹ in the tumour. This reaction has been studied extensively in biological frameworks, which include in vitro, in vivo and clinical trials. Much effort has been expended on glioblastoma multiforme (GBM), for which epithermal NCT with BPA-F was found to achieve results comparable to or better than standard therapies.

**Abbreviations:** BPA-F = 2-p-boronophenylalanine fructose, GBM = Glioblastoma multiforme and LET = Linear energy transfer.

**Related Articles:** Fast neutron therapy (FNT), Boron neutron capture, Linear energy transfer

Neutron therapy
(Radiotherapy) In neutron therapy, fast neutron beams of typical energy 20–50 MeV are used to treat deep seated tumours. The beams are usually produced using high energy proton beams impinging on a beryllium target and may be shaped using a multileaf collimator (MLC). Compared with x-rays, neutrons have the same exponential attenuation with depth, being also uncharged particles, but the ionisation density from a neutron is higher than the same exponential attenuation with depth, being also uncharged particles, but the ionisation density from a neutron is higher than that of x-rays.

**Abbreviations:** LET = Linear energy transfer, MLC = Multileaf collimator and RBE = Relative biological effect.

**Related Articles:** Charged particle therapy, Hadron therapy, Ion therapy, Proton therapy, Heavy particle beams

Neutrons
(General) Neutrons are subatomic particles which were discovered in 1932 by James Chadwick. They are composed of two down quarks and one up quark held together by the strong nuclear force. Free neutrons are unstable, decaying with a half-life of 10.3 min into a proton, electron, and antineutrino via beta-minus decay. Along with protons, neutrons form the constituent parts of the atomic nucleus.

**Medical Applications:** Boron neutron capture therapy – In this form of radiotherapy, the interaction between an external neutron beam and boron (injected intravenously) is used.

External beam radiotherapy – Neutron treatment beams have been used to treat hypoxic tumours (neutrons have a reduced oxygen enhancement ratio compared to x-ray photons), but due to unacceptable late normal tissue damage this practice has largely been abandoned.

Radiation hazard – Additional shielding is required for high energy clinical linear accelerators (>10 MeV) as here neutrons are produced via both electron–neutron and x-ray–neutron interactions.

**Related Articles:** Atom, Half-life, Gyromagnetic ratio, Boron neutron capture, Oxygen enhancement ratio, Radiation shielding

Nickel
(General) Ni

| Symbol | Ni |
| Element category | Transition metal |
| Mass number A | 58 |
| Atomic number Z | 28 |
| Atomic weight | 58.694 g/mol |
| Electronic configuration | 1s² 2s² 2p⁶ 3s² 3p⁶ 4s² 3d⁸ |
| Melting point | 1728 K |
| Boiling point | 3186 K |
| Density near room temperature | 8.908 g/cm³ |

**History:** Nickel was first extracted from nickel arsenide in 1751, by Baron Axel Fredrik Cronstedt. It has been found in bronze dating from up to 3500 BC, although it was not identified until Cronstedt’s discovery. Since 1859, nickel has been used in pure form or as an alloy to make coins.

**Isotopes of Nickel:** Five stable isotopes of nickel are found in nature. The most abundant is ⁶⁰Ni, with 68.1% natural abundance, followed by ⁶⁴Ni (26.2%), ⁶²Ni (3.6%), and ⁶¹Ni and ⁶⁵Ni, both at around 1%. Eighteen synthetic radioactive isotopes are known, which have half-lives ranging from 76,000 years (⁶⁰Ni) to 110 ms (⁶⁷Ni).

Nickel is ferromagnetic and exhibits negative magnetostriiction: slight contraction in the presence of a magnetic field.

**Medical Applications:** Magnetic resonance imaging – Nickel, along with aluminium and cobalt, forms part of an alloy known as alnico, which is used for manufacturing permanent magnets used in MRI. Although superconducting magnets are now more widely used, permanent magnets may be advantageous in some situations as they require no cooling and have small fringe fields. Permanent magnets are often used in ‘open’ magnet configurations, which are patient-friendly and enable interventional MRI.

Nickel, in the form of NiCl, is also used as a dopant in MR phantoms to adjust the relaxation times of the phantom contents to more closely resemble those of biological tissue. The relaxation times of NiCl are largely independent of temperature and operating frequency.

**Related Articles:** Magnetic resonance imaging (MRI), Magnet(s), Permanent
Nitrogen
(Diagnostic Radiology) See Brightness

Nitrogen (General)

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<th>Symbol</th>
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<td>Boiling point</td>
<td>77.36 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
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</tr>
</tbody>
</table>

History: The discovery of nitrogen is formally credited to Daniel Rutherford, who published a report in 1772 describing a component of air in which combustion was not supported.

Isotopes of Nitrogen: 99.634% of naturally occurring nitrogen is stable ¹⁴N. The remaining 0.366% is stable ¹⁵N. There also exist 10 synthetic radioactive isotopes, with ¹⁵N the only isotope with a half-life greater than a few seconds (the half-life of ¹⁴N is 10 min).

Medical Applications: Cryogen – Nitrogen is predominantly used in liquid form as a coolant, for example as an alternative to liquid helium in MRI cooling systems. It can also be used as a preservative for storing biological tissue.

PET Imaging Agent – ¹⁵N-ammonia is used as a radiopharmaceutical in PET imaging studies. It has a high first pass extraction rate into the myocardium, making it useful for myocardial perfusion imaging.

Related Articles: Cryogen, PET (positron emission tomography), PET clinical applications, Perfusion imaging

Nitrogen, liquid
(General) See Nitrogen

NMR (nuclear magnetic resonance)
(Magnetic Resonance) See Nuclear magnetic resonance (NMR)

Noise
(Nuclear Medicine) In nuclear medicine noise refers to the background contribution in images, that is corrupted signal without any true image information. The background contribution is a by-product of other activities, for example electric noise in detectors and background radiation. The noise contribution in an image can be expressed in a signal-to-noise ratio, SNR. Images can be smoothed in the image post processing in order to raise the signal-to-noise ratio.

Related Article: SNR (signal-to-noise ratio)

Noise equivalent quanta (NEQ)
(Diagnostic Radiology) Noise equivalent quanta (NEQ) describes the number of photons (quanta) which are incident on a detector as derived from the detected signal-to-noise ratio (SNR). In the case of an ideal detector the NEQ is proportional to the square of the measured SNR, that is the measurement follows photon counting statistics. For a non-ideal detector, the NEQ is defined using the same assumption and hence an equivalent number of incident photons can be defined assuming a perfect detector.

The NEQ is used in combination with the detective quantum efficiency (DQE) to quantify the intrinsic efficiency of a radiographic imaging system such as an x-ray flat panel detector. NEQ provides a quantitative measure of the quality of the image. This can then be used to infer, for example the minimum x-ray dose to the patient to provide images of satisfactory quality in an x-ray radiograph.

For an ideal photon counting detector the number of incident photons (Ninc) is proportional to the SNR by the following (Equation N.1):

\[ N_{\text{inc}} \propto \text{SNR}^2 \]  

(N.1)

In a real measuring system the detector is not perfect and not every photon incident on it is converted to a measured count. This reduces the SNR in comparison to an ideal detector in which every photon is measured. The measured SNR of a real, non-ideal detector is denoted by SNR_{meas}. Furthermore, there are additional noise sources within the detector which add to the noise associated with measured signal further reducing SNR_{meas}. The measured non-ideal SNR can thus be related to the equivalent number of quanta that would be incident on the detector by (Equation N.2)

\[ NEQ \propto \text{SNR}_{\text{meas}}^2 \]  

(N.2)

For the case of a real medical imaging detector both the signal measured by the detector and the inherent noise generated become spatial frequency dependent quantities. Thus, the NEQ must also be frequency dependent. The signal output of an imaging device is related to the input signal via the modulation transfer function (MTF(f)). This describes how well the imaging device amplifies the input signal over the various spatial frequencies. The frequency dependence in the noise of the device is denoted by the noise power spectrum (NPS(f)). This describes the spatial frequency distribution of the noise inherent to the device. The frequency dependent NEQ can now be defined as (Equation N.3)

\[ \text{NEQ}(f) = \frac{q^2 \text{MTF}(f)^2}{\text{NPS}(f)} \]  

(N.3)

where \( q \) is the number of incident photons per unit area (the average uniform input). This has replaced the term \( N_{\text{inc}} \) in Equation N.1 due to NPS being a measure of the noise power per unit area. MTF(f) and NPS(f) are both linearised functions with respect to input intensity.

The interpretation of the frequency dependent noise equivalent quanta (NEQ(f)) is very similar to that for standard NEQ. NEQ(f) tells you the number of Poisson distributed quanta that would produce the same SNR in an ideal detector at a particular spatial frequency component.

Related Articles: Noise power spectrum (NPS), Detective quantum efficiency (DQE), Modulation transfer function (MTF), Signal-to-noise ratio (SNR)


Noise power spectrum (NPS)
(Diagnostic Radiology) The noise power spectrum (NPS) is the power of noise, contained in a two-dimensional (2D) spatial frequency interval, as a function of the 2D frequency.

The noise power spectrum or Wiener spectrum is used to calculate the magnitude and frequency of noise in medical imaging...
systems. Although applicable to all imaging analysis it is used mostly in diagnostic x-ray imaging. A measurement of the NPS is also needed to determine the noise equivalent quanta (NEQ) and the detective quantum efficiency (DQE) of an imaging system. These quantities allow quantitative evaluation and comparison of imaging system efficiency.

There are many sources of noise in an x-ray imaging system. In a perfect detector, the number of photons measured per unit area will vary randomly due to small fluctuations in the number of incident photons on the detector, or quantum noise. However, in real systems, other forms of noise will also contribute to the final image, for example imperfect detection of incident photons, and internally generated system noise. By measuring the NPS of a system both the random and non-random noise is quantified.

In 2003, the International Electrotechnical Commission (IEC) published an international standard for the measurement of the DQE, which included a detailed description of NPS measurement, on which the following description is based. Figure N.6 depicts the basic calculation stages. As the NPS describes the noise in terms of the magnitude of frequency components, Fourier analysis is used to calculate the spectrum.

The NPS is measured by first taking a flat field image, an image with no test object in the beam with a standard spectrum and geometry. A flat field image appears dark grey and uniform to the naked eye, for with no object signal the image contains only noise components. Several exposures should be taken as it is recommended that there be 4 million independent pixel signals for measurement. It should be noted that the taking of several images assumes that there is no noise component that changes with time, that is the noise components are either completely random or temporally reproducible and thus do not change over time. All images must be linearised to convert the signal intensity to units of quanta absorbed per area. This is especially important if the DQE will be measured from the NPS as linearising the image negates the need for a further gain factor in the DQE measurement. All measurements are taken from a central region of interest (ROI) of area $125 \times 125$ mm. This analysis ROI is then ‘detrended’ by subtracting a second-order polynomial from the image signal. This detrending is used to remove some non-random image intensity fluctuations such as the intensity gradient caused by the anode heel effect; however it may also remove some noise components that are wanted be measured, such as structured noise, and some investigators withhold this step.

**Figure N.6** (See colour insert.) Measurement of NPS. (Graphs courtesy of KCARE, UK www.kcare.co.uk).
The analysis region is then broken up further into half-overlapping regions of interests made of 256 x 256 pixels. For each ROI the 2D Fourier transform is calculated and then the modulus squared is taken, the final NPS is found by averaging all the 2D Fourier transforms. Mathematically for a 2D, discrete (pixellated) image the NPS is defined as

\[ \text{NPS}(u_x, v_y) = \frac{1}{M} \sum_{i=1}^{M} \left( I(x_i, y_j) - S(x_i, y_j) \right) \left| e^{-2\pi i (u_x x_i + v_y y_j)} \right|^2 \]

where
- \( u_x, v_y \) are frequency components in each axis direction in Fourier space
- \( \Delta x, \Delta y \) are the pixel spacing
- \( N_x, N_y \) are the number of pixels in the individual ROIs
- \( M \) is the number of ROIs or ensemble averages
- \( I(x_i, y_j) \) is the linearised data
- \( S(x_i, y_j) \) is the 2D polynomial

The NPS can be shown as either a 3D graph, or as a 2D line graph depicting the horizontal and vertical NPS. The horizontal and vertical NPS are found by averaging NPS at each frequency for ±7 points either side of the axes.

**Related Articles:** Modulation transfer function (MTF), Noise equivalent quanta (NEQ), Detective quantum efficiency (DQE)


**Nominal output**

(General) See Nominal value

**Nominal value**

(General) Nominal value is the stated value of a physical quantity or a property of a device, under normal conditions. The actual value may differ from the nominal for an absolute value or percentage declared by the tolerances. For example, the nominal value of line voltage in Europe is 230V, with tolerance ±10%. Resistors are produced with nominal values taken from Renard series of preferred numbers and defined tolerances (20%, 10%, 5%, etc.).

**Non-coplanar beams**

(Radiotherapy) Typically the fields used for radiotherapy treatment are delivered in a single plane around the patient. However, on occasion it is beneficial to treat with beams incident outside a single plane. This can be achieved through a combination of linac movement (gantry rotation) and couch movement (couch rotation). Non-coplanar beams are most commonly used for brain and head and neck treatments where the target is surrounded by critical structures. An illustration of a non-coplanar beam arrangement is given (Figure N.7).

**Non-ionising radiation**

(Radiation Protection) Non-ionising radiation refers to any part of the electromagnetic (EM) spectrum that cannot induce ionisations in matter. In other words, unlike ionising radiation, non-ionising radiation does not have sufficient energy to interact directly with sub-atomic electrons. Types of non-ionising radiation that are used clinically include radio-frequency fields (in MRI) and ultraviolet (for dermatological therapy).

The minimum energy for ionisation — that is, giving electrons enough energy to leave their orbits — is around 12eV, which is equivalent to the energy in an electromagnetic wave of wavelength approximately 100nm; this is at the far end of the ultraviolet range of the electromagnetic spectrum. Below the 12eV threshold for atomic ionisation, the radiation may still have enough energy to break molecular bonds in human tissues. The threshold for this effect may be as low as 4eV. This threshold lies in the UVB range of the spectrum.

‘Non-ionising radiation’ does not mean ‘non-hazardous’. Despite their inability to invoke ionisation events in matter, non-ionising radiation can have a damaging effect on tissue. Ultraviolet radiation can cause skin erythema, intense light or near-infrared radiation can burn the cornea or retina of the eye, whilst high levels of lower-frequency EM radiation can cause tissue heating at depth within the body. For example, the rapidly varying radio-frequency waves that re-align the magnetic dipole of water molecules leading to a signal detected in an MRI scanner, may also cause heating in the body tissues of a patient.

The range of effects of non-ionising radiation on the tissues of the human body is still not fully understood. The International Commission on Non-Ionizing Radiation (ICNIRP) has published guidance in this field, including proposed guideline limits for occupational and public exposure (see Further Reading).

**Related Articles:** Ultraviolet radiation (UV), Infrared radiation (IR)


**Non-linear dose–response curve**

(Radiation Protection) The non-linear dose–response curve is just one example of a dose–response curve, together with non-threshold dose–response curve, linear response curve, etc., which may be used either separately or in combination as models to...
describe the response of the human body to exposure to various types on ionising radiation from both internal and external exposure, and at high and low doses and dose-rates.

The non-linear dose–response does not assume that there is a linear response between exposure at low doses/dose-rates and exposure at high doses/dose-rates.

**Related Articles:** Dose–response curve, Linear dose–response curve

### Non-linear propagation

(Ultrasound) Non-linear propagation of a sound wave can be observed as a distortion of the wave as it propagates, as shown in Figure N.8. A propagating wave is usually described by the wave equation, but the wave equation cannot explain this behaviour. This is due to the fact that as the equation is derived, a number of non-linear terms are neglected based on the assumption that the pressure perturbations are small. In diagnostic ultrasound however, the pressure amplitudes are on the order of a magnitude higher than the atmospheric pressure, and therefore the non-linear terms will not be negligible. In practical terms this has two effects when a propagating wave is considered:

1. **Description of Figure N.8** – the top panel shows the wave as it appears close to the transducer. The next two panels show how the waveform progressively is distorted as it propagates, and the last panel how it appears after energy has been attenuated after a distance, and the amplitude is not sufficient to produce the non-linear effect.

2. **Convection:** First, there is a convection effect. For large acoustic pressures, the particle velocity will also be considerable, and thereby also the ratio \( v/c \) (particle velocity over the sound speed). As a consequence the wave tends to migrate as \( c+v \). This means that in the compression phases, when the particle velocity is at its maximum and in the propagation direction, the sound speed is higher than in the rarefaction phases, where the particle velocity is at its maximum in the opposite direction.

3. **Non-Linear Compressibility:** Second, with increased pressure, the stiffness and bulk modulus of a fluid tend to increase. As the sound speed is proportional to the square root of the bulk modulus, the sound speed will also increase with pressure due to this effect.

4. **Practical Effects:** These effects manifest themselves in the way that the compressions (where there is high pressure) travel faster and ‘catches up’ with the rarefactions (where the pressure is low). The effect is a shocked wave front as depicted in Figure N.6, middle panels.

The shocked waveform has a very high harmonic content due to the asymmetric waveform, and as attenuation is frequency dependent, this counteracts the build-up of harmonics. After propagating a longer distance, the energy transfer to higher harmonics is insufficient, and the wave again appears as a sinusoidal wave, but with lower amplitude (Figure N.6, bottom panel).

5. **Related Article:** Harmonic imaging

### Non-linearity parameter

(Ultrasound) The non-linearity parameter (or the \( B/A \) value) indicates how a medium supports the build-up of non-linear waves. A high intensity wave will generate changes in the speed of sound. The perturbed speed of sound can be written as

\[
c(t) = c_0 \left(1 + \frac{B}{2A} u(t) \right)^{\frac{1}{2}}
\]

where
- \( c_0 \) is the speed valid for small pressure variations
- \( u(t) \) the particle speed
- \( B/A \) is a material dependent parameter

For water \( B/A = 5.2 \) (at 30°C and atmospheric pressure), whereas for air it is 0.2. For tissues \( B/A \) falls in the range of 5–11.

Also used is the non-linear coefficient \( \beta \), which is \( \beta = 1 + B/2A \).

### Non-paralysable counting system

(Radiation Protection) The dead time delay in a detector is a minimum time interval which must separate two events to be measured (registered) as two separate pulses.

In non-paralysable counting system the events that occur during the dead time period are lost, so that with an increasing event rate the detector will reach a saturation rate equal to the increase of the dead time.

If we assume that \( N \) is a true interaction rate and \( M \) is a measured counts rate for the detection system with dead time \( \tau \), the difference between a true and measured rate is equal to

\[
N - M = N \times M \times \tau
\]

where \( M \times \tau \) is the fraction of all time when the detector is ‘dead’.

Then the true interaction rate \( N \) can be calculated from the following equation:

\[
\frac{N}{1 - M \times \tau} = M
\]

**Related Articles:** Dead time losses, Paralysable counting system


### Non-scattering grid

(Diagnostic Radiology) Non-scattering grid is also known as anti-scattering grid. See article Grid, Bucky.

**Related Articles:** Grid, Bucky

### Non-screen film

(Diagnostic Radiology) Non-screen film is film that is exposed without the use of intensifying screens. The film is placed in black envelope and in the past had been used for some high-resolution...
radiographs. Nowadays it is not used for human radiography because high exposures are required. It is useful for some applications such as specimen radiography.

The advantage of non-screen film is that it can produce an image with high visibility of detail because the blurring by the intensifying screen is eliminated.

It can be useful in the measurement of geometrical parameters in the quality controls of x-ray and radiotherapy equipments.

Non-stochastic effects
(Radiation Protection) There are two types of biological effects (Bioeffects) of ionising radiation on human tissues categorised by the risk of the effects being observed. These two categories are stochastic and non-stochastic effects.

Non-stochastic effects result from cell killing due to radiation exposure, and they occur in all persons exposed to a relatively high radiation dose above a threshold.

The cells in many tissues and organs of the body are continuously being replaced. However, the balance between loss and replacement rates may be affected by many factors including exposure to radiation. The result of a net loss of cells may be the failure of function of the tissue or organ. Gross reductions in the number of healthy cells in vital tissues may lead to death. For each tissue type there is an absorbed dose threshold level below which the probability of harm is very low. Above this threshold, pathological conditions will be observed in nearly all the irradiated population and their severity will be dose-dependent. Below the threshold, it is assumed that the body is able to replace the cells at a faster rate than they are being killed – hence no damage is detected. The following diagram represents the assumed relationship between non-stochastic effects and radiation dose received (Figure N.9).

Examples of these non-stochastic effects range from eye cataracts, sterility, skin-reddening (erythema), and effects on a foetus, all of which occur at moderate exposures, to the symptoms of acute radiation syndrome such as nausea and vomiting at exposures near lethal levels. Because of the onset of clinically observable effects of a deterministic nature, the International Commission on Radiological Protection now calls them ‘deterministic effects’.

Related Articles: Bioeffects, Deterministic effect, Stochastic effect


Non-threshold dose–response curve
(Radiation Protection) The curve that describes the dose–response relationship for excess risk of cancer following exposure to ionising radiation at low dose and/or low dose rates with the assumption that there is no threshold value below which there is no risk.

Presently, for the purpose of radiological protection, the assumption is made that the underlying dose–response relationship is linear-quadratic with no threshold.

Related Article: LNT model

Non-transparent
(Diagnostic Radiology) See Opacity

Non-uniform activity distribution
(Nuclear Medicine) A non-uniform tracer distribution in an organ or a compartment. Non-uniform distribution of tracers allows imaging of biological and pathological processes, for example FDG-compound is accumulated in regions with a high glucose uptake which can be an indication of tumour cells.

When calculating the radiation dose to patients using the MIRD formalism one assumes a uniform activity distribution inside an organ. This is an obvious flaw to the MIRD formalism since the activity is not uniformly distributed. Future research with high-resolution detectors and elaborate Monte Carlo simulations will allow for better determination of the radiation dose.

Non-uniformity
(Magnetic Resonance) Intensity non-uniformity is the smooth intensity variation often evident in MR images, even those of uniform media (Figure N.10).

Image non-uniformity in MRI is caused primarily by the following factors:

1. Transmit/receive coil RF non-uniformity
   This is perhaps the most significant determinant. In general, transmit coils are required to have very good uniformity so that transmission flip angles are constant throughout the volume to be imaged. For receive coils, good uniformity is generally less of a priority than good SNR. For this reason image non-uniformity is generally more dependent on RF reception rather than RF transmission.

2. Nyquist limit filtration
   To prevent aliasing, active filters within the RF receiver strongly attenuate all received frequencies above the

**FIGURE N.10** Image of a uniform test object. If the image represented the test object faithfully, all test object pixels would be of uniform intensity. In reality the voxels towards the centre of the test object are less intense than those at the periphery: this is image non-uniformity.
Nyquist limit. Such filtration can introduce non-uniformities to the image profile in the frequency encoding direction on older MR systems.

3. **Eddy currents**

Eddy currents (interaction between the metal magnet housing and the rapidly switched RF gradient fields) can cause image non-uniformities.

4. **Pulse sequence parameters**

Some pulse sequence parameters can adversely affect image uniformity. For example, if the inter-slice spacing is too low, cross talk between adjacent slices can sometimes affect image uniformity. Slice profile and slice width are also important parameters.

5. **Dielectric resonance effects**

If high frequency RF pulses have wavelengths of the order of the dimensions of the human body, dielectric resonant interactions between the RF fields and the body can occur, causing intensity non-uniformity.

**Measuring Image Uniformity:** Quantitative values for uniformity can be extracted from images of a uniform (flood field) test object.

The Institute of Physics and Engineering in Medicine (IPEM) recommend the use of intensity profiles such as that shown in Figure N.11. Their method is as follows:

1. On the test object image, draw a roughly central profile parallel to the frequency encoding direction, avoiding any artefacts.
2. Find the modal value of the 100 pixels at the centre of this profile.
3. Calculate the fractional uniformity: the fraction of the profile which lies within ±10% of the modal value.
4. Repeat for 10 profiles and take a mean to reduce the effect of noise.
5. Repeat steps 1–4 for profiles in the phase encoding direction.

**Results:** IPEM suggest that image uniformity should typically lie between 0.6 and 1.0.


---

**Normal database**

(Nuclear Medicine) The gender-specific normal database includes information from a number of patients with confirmed low likelihood of coronary artery disease. The gender-specific normal limits are derived from the mean of normal uptake in a particular region and its standard deviation. The criteria for abnormality of a specific region of the myocardium are based upon a comparison between regions for healthy and corresponding regions for diseased patients. When the circumferential profiles for individual patients match those normal limits, the area of abnormal pixels that falls below them and the relative depth of severity can be calculated. If the myocardial perfusion images are not attenuation corrected then there is no simple relationship between tracer uptake and the number of counts in each pixel. Image scaling is therefore necessary often relative to the maximum pixel counts.

It is necessary to develop limits for normality and criteria for abnormality for the different protocols. The mean and standard deviation of the gender-specific database for the normal response were calculated based on a limited number of male and female subjects with a <5% likelihood of having CAD. The normal limits and criteria for detection of perfusion abnormalities were performed based on a number of male and female patients undergoing 99mTc-Sestamibi SPECT and coronary angiography. The normal limits of a gender-specific normal database in different regions of the myocardium optimise the balance of sensitivity and specificity in the detection of perfusion defects and the assessments of their reversibility, extent and severity.

Normal databases are used in other imaging applications such as brain perfusion imaging for statistical parametric mapping.


**Normal distribution**

(General) The normal distribution is synonymous with the Gaussian distribution, and is considered to be the most important continuous probability distribution, since it describes the behaviour of many random variables of interest. It can also be used to approximate other more complicated distributions.

If a random variable $X$ follows a Gaussian distribution with mean $\mu$ and variance $\sigma^2$, we write:

$$X \sim N(\mu, \sigma^2)$$

The equivalent probability density function is described as follows:

$$f(x) = \frac{1}{\sigma\sqrt{2\pi}} \exp\left(-\frac{1}{2} \left(\frac{x-\mu}{\sigma}\right)^2\right)$$

The normal distribution has the following properties, which are illustrated in Figure N.12:

- It is symmetric about the point $x=\mu$
- It has a characteristic ‘bell’ shape
- The width of the curve is described by the standard deviation $\sigma$
- The points $x=\mu \pm \sigma$ are inflection points where $d^2f/dx^2 = 0$
Normal organ dose tolerance

(Radiotherapy) Radiation treatment inevitably affects normal tissue and so may cause radiation-induced adverse effects. In radiotherapy, it is generally the case that the total dose that can be tolerated depends on the volume of tissue irradiated – the dose–volume effect. Additionally, the tissue architecture is thought to be important in determining the tolerance dose for partial organ irradiation. For further information, see the article Tolerance.

Related Article: Tolerance

Normal speed tube

(Diagnostic Radiology) The speed of x-ray tube anode rotation depends on the frequency of the current supplying its stator. Usually normal speed is considered for stator supply with 50 Hz (or 60 Hz in the United States), allowing rotation with approximately 3000 rpm (in fact due to the mechanical slipping of the rotor this speed is around 2800 rpm).

Related Articles: Anode, Rotation anode, Anode rotational speed, Anode acceleration


Normal tissue complication probability

(Radiotherapy) The relationship between dose and normal tissue complication probability (NTCP) is shown in Figure N.13. It has a sigmoid (S) shape with the probability of tumour control tending to zero as the dose tends to zero and tending to 100% at very large doses.

Several radiobiological models have been proposed that relate biological effect to volume and dose distribution (dose–volume models). Probably the most widely used to date is that of Lyman (1985) in which it is assumed that NTCP is a function of both the fractional volume irradiated and the absorbed dose received by the volume. Many other factors are involved, particularly the

\[
P(\mu - \sigma < X \leq \mu + \sigma) = 68.3% \\
P(\mu - 2\sigma < X \leq \mu + 2\sigma) = 95.4% \\
P(\mu - 3\sigma < X \leq \mu + 3\sigma) = 99.7%
\]

Related Article: Gaussian distribution

FIGURE N.12 Gaussian distribution or normal distribution for mean 3 and various values of the standard deviation.

The dose dependence is represented by the integral of a normal distribution which is just one of several possible representations of a sigmoid curve, Equation N.5:

\[
NTCP = \frac{1}{\sqrt{2\pi}} \int_{-\infty}^{t} e^{-(x/\sigma)^2} dx
\]

where

\[
t = \frac{D - TD_{50}(V)}{\sigma(V)}
\]

The model utilises the \(TD_{50}(V)\) (\(TD_{50}\) in Equation N.5), the dose that would result in 50% incidence of complications in a population after 5 years. The sigmoid curve is completely defined by the mean \(TD_{50}(V)\) and the standard deviation \(\sigma(V)\) which for this model was approximated as shown in Equation N.6 where \(m\) is a fitted parameter:

\[
\sigma(V) = m \times TD_{50}(V)
\]

Lyman gave estimates of \(TD_{50}(V)\), \(n\), and \(m\) based on the clinical data available at the time. The complication end points were not specified. Burman et al. (1991) obtained Lyman model parameters for the clinical tolerance data compiled by Emami et al. (1991). In the Lyman model, when \(n \rightarrow 1\) there is a large volume effect and when \(n \rightarrow 0\) the volume effect is small. Figures N.14 and N.15 show the three-dimensional (3D) surfaces obtained from the Lyman model using the data of Burman et al. for lung (\(n = 0.87\)) and rectum (\(n = 0.12\)) respectively. It must be remembered that the Emami data were largely the result of pooled clinical experience and since most therapies are by definition designed to generate small NTCPs (on the order of 5% or less, see also the article Therapeutic effect), doubt has been expressed over the credibility of those points of the sigmoid response curve corresponding to much larger NTCP values.
et al. (1993) have described the biological data needed as input for determining the tolerance dose for partial organ irradiation (see example, the tissue architecture is thought to be important in the underlying mechanisms of normal tissue complications. For partial organ irradiation may result from a better understanding of NTCP still utilises empirical models that are limited in that they are not obvious. A comprehensive review of NTCP models may be found in Mayles et al. (2007, Chapter 36).

**Abbreviations:** DVH = Dose–volume histogram, LKB = Lyman–Kutcher–Burman model, NTCP = Normal tissue complication probability and \(TD_{50/5} = \) The dose that would result in 50% incidence of complications in a population after 5 years.

**Related Articles:** Dose–response model, Dose–volume histogram, Fractionation, Parallel organ, Radiobiological models, Serial organ, Sigmoid dose–response curve, Therapeutic effect, Tolerance.


**Normal tissue dose** *(Radiotherapy)* Radiation treatment inevitably affects normal tissues and the dose they receive is a contributing factor to the induction of adverse effects. For further details on the response of normal tissue to radiation see the articles on *Adverse effects, Dose–response model and Tolerance.*

**Related Articles:** Adverse effect, Dose–response model, Tolerance

**Normal tissue dose–response** *(Radiotherapy)* Radiation treatment inevitably affects normal tissues and their response depends on a number of factors including the dose received, the volume of tissue irradiated, the tissue architecture and the fractionation regime. For further details on the response of normal tissue to radiation see the articles on *Adverse...
effects, Dose–response model, Normal tissue complication probability and Tolerance.

Related Articles: Adverse effect, Dose–response model, Fractionation, Normal tissue complication probability, Tolerance

Normal tissue reaction
(Radiotherapy) Radiation treatment inevitably affects normal tissue and so may cause radiation-induced adverse effects. For further details on radiation-induced normal tissue reaction, see the articles on Adverse effects, Dose–response model, Normal tissue complication probability and Tolerance.

Related Articles: Adverse effect, Dose–response model, Normal tissue complication probability, Tolerance

Normal tissue toxicity
(Radiotherapy) Radiation treatment inevitably affects normal tissue and so may cause radiation-induced toxicity. For further details see the articles on Adverse effects and Tolerance.

Related Articles: Adverse effect, Tolerance

Normalisation point
(Radiotherapy) The normalisation point is a specific point to which the 3D matrix of dose values calculated by the TPS is normalised, that is a relative dose of 100% is assigned to this point.

There are two principally different ways of normalisation point selection:

1. The normalisation point position depends on the details of the treatment plan or on the calculated dose distribution.
   The isocentre, the intersection of several beam axes, the minimum or maximum dose value in a slice or in a specified volume (entire volume, PTV, OAR, ...) are examples of such normalisation points. When two or more competing treatment plans are then compared for a given patient, they are in general normalised to different normalisation points.

2. The normalisation point position is independent of the treatment plan.
   In this approach, a fixed normalisation point is defined in a clinically relevant location, the ‘ICRU reference point’ being an example. The ICRU introduces in its reports the concept of ‘ICRU reference point’ to which 100% relative dose should be assigned. The ICRU point should be located firstly at the centre, or in the central part, of the PTV, and secondly on or near the central axis of the beam(s). In this approach, competing treatment plans of a particular patient are normalised to the same point in the patient body.

Abbreviations: ICRU = International Commission on Radiation Units and Measurements, OAR = Organ at risk, PTV = Planning target volume and TPS = Treatment planning system.

Related Articles: Treatment planning system, Isodose curve, Critical structures, Organ at risk, Planning target volume, Prescribed dose


Normalised treatment dose (NTD)
(Radiotherapy) This refers to the treatment dose at the normalisation point in a treatment plan. Please see the article Normalisation point for further information.

Related Articles: Normalisation point, Treatment plan normalisation

Notification
(Radiation Protection) National laws and regulations include the duty to notify to the national regulatory authority the intention to carry out a practice (see below and Further Reading). A person, or their legal representative, intending to carry out any practice, as defined by the regulatory authority, or other actions specified by the regulatory authority, shall notify the regulatory authority and supply the details and documents specified by the regulatory authority.

A practice is any human activity that introduces additional sources of exposure or exposure pathways or extends exposure to additional people or modifies the network of exposure pathways from existing sources, so as to increase the exposure or the likelihood of exposure of people or the number of people exposed.

Related Article: Regulatory authority


NPL
(General) The National Physical Laboratory at Teddington is the United Kingdom’s National Measurement Institute.

Hyperlink: www.npl.co.uk

NRA
(Radiation Protection) National regulatory authority. For further information, see Regulatory authority.

Related Article: Regulatory authority

NRC
(Radiation Protection) See Nuclear Regulatory Commission (NRC).

Nuclear activation analysis
(Nuclear Medicine) Activation analysis depends on the observation that almost every element can be converted into a radioactive isotope by bombardment with neutrons or charged particles. The induced radioactivity is highly specific of the radionuclides contributing to it. By studying the induced radioactivity, the constituent elements in the sample may be identified and quantitatively estimated. This apparently round-about procedure is, for more than half of the chemical elements, the most sensitive analytical technique available. The gain in sensitivity by comparison with other methods can be on the order of 10³.

Related Articles: Induced radioactivity, Radioactivity, Cyclotron, Nuclear reactor


Nuclear binding energy
(General) Nuclear binding energy is the energy to separate a nucleus into its unbound neutrons and protons.

The mass of a nucleus is less than the sum of the masses of its constituent protons and neutrons. The difference in mass (mass
defect) is the binding energy in accordance with Einstein’s formula \( E = \Delta mc^2 \), where \( E \) is the binding energy, \( \Delta m \) is the mass defect and \( c \) is the velocity of light.

Nuclear binding energy is derived from the nuclear force.

The mass defect per nucleon varies with the mass of the nuclide as shown in Figure N.16.

**Related Articles:** Nuclear force, Nucleus, Nucleon

**Nuclear chain reaction**

(General) A self-propagating nuclear reaction, in which the occurrence of a nuclear reaction causes further nuclear reactions. The term is used predominantly to describe nuclear fission reactions, where a neutron is absorbed by a heavy fissile nucleus, causing the nucleus to split into a number of fission fragments and releasing energy (usually in the high MeV range). In a nuclear chain reaction, neutrons produced through the fission of one fissile nucleus go on to initiate further nuclear reactions by absorption in the surrounding material.

**Uses in Medicine:** Nuclear medicine – Nuclear chain reactions in a nuclear reactor play a part in the production of many radionuclides used in nuclear medicine. The most common nuclear medicine radionuclide, \(^{99}\text{mTc} \), is generated by the decay of unstable molybdenum, which can itself be produced through the fission of uranium nuclei in a nuclear reactor, generating neutrons which take part in further reactions:

\[
^{235}\text{U} + n \rightarrow ^{236}\text{U}^* \rightarrow ^{134}\text{Sn} + ^{99}\text{Mo} + 3n + \gamma
\]

**Nuclear emissions**

(Radiation Protection) ‘Nuclear emissions’ is a collective term covering all particulate and electromagnetic radiation emitted during radioactive decay or other nuclear event. The most common particles involved are alpha (\(\alpha\)) and beta (\(\beta\)) particles, neutrons, protons and neutrinos. The term ‘gamma rays’ is used for the high energy electromagnetic radiation emitted from a nucleus during radioactive decay. X-rays and electrons can be emitted from the atom involved as a consequence of interactions between the nucleus or emitted particle and the electrons in the shells surrounding the nucleus. In interactions of very high energy particles with the nucleus other sub-atomic particles can be emitted.

The term ‘nuclear emissions’ is sometimes loosely used instead of the term ‘nuclear discharges’ – the release of radionuclides into the environment (sea, air, river, etc.) from a nuclear facility.

**Related Articles:** Alpha particle, Beta particle, Neutrinos, Nuclides, Radioactive decay

**Nuclear fission**

(General) Nuclear fission is when a nucleus splits into two separate nuclei accompanied by the emission of particles and gamma rays. In most cases the particles emitted are neutrons but in some instances either alpha (\(\alpha\)) particles or light nuclei are emitted. Tertiary fission (three fission fragments) occurs very rarely.

Nuclear fission can occur spontaneously in high mass radioactive nuclides. An example is californium-252, which has a half-life of 2.6 years. It decays by alpha decay in 97% of transformations and by spontaneous fission in 3%.

Nuclear fission is normally induced by a neutron and an example is the induced fission of uranium-235 in a nuclear reactor. Various fragment pairs may be result but an example is

\[
\frac{235}{92} \text{U} + \frac{1}{0} n \xrightarrow{\text{fission}} \frac{137}{55} \text{Cs} + \frac{97}{37} \text{Rb} + \frac{1}{0} n + \frac{1}{0} n + \text{Energy (about 200MeV)}
\]

The sum of the masses of the product nuclei must be 234, whilst the sum of the atomic numbers of the products must equal the atomic number of uranium for the equation to balance. Figure N.17 illustrates the relative amounts of the products of the fission of U-235 – one fission fragment from each of the peaks. Over 60 primary products have been detected. Most of the primary products are themselves radioactive and decay into a wide range of nuclides.

**Related Articles:** Alpha particle, Nuclear reactor, Radioactive decay

**Nuclear forces**

(General) The nuclear force acts between the nucleons (nucleon–nucleon interaction) in the nucleus of the atom and holds the nucleons (protons and neutrons) together, being much stronger than the electrostatic repulsion between protons.

![FIGURE N.17](image_url) Distribution of the yield of fission products from U-235 as a function of mass number (data smoothed). Note log scale for the ordinate.
The nuclear force, also referred to as the strong force, strong nuclear force or residual strong force, is one of the four basic forces of nature – the others are gravity, the electromagnetic force and the weak nuclear force.

The force is short range, essentially acting only within the nucleus, and is mediated by mesons.

A full explanation and current knowledge of the nuclear force is complex as the nucleons are themselves composed of quarks with gluons involved as the carriers of the force. Quantum chromodynamics is a theory of the interaction between quarks and gluons.

**Related Article:** Nucleons

**Nuclear instability**

*(General)* Nuclides which do not have a stable combination of neutrons and protons (the nucleons) transform spontaneously into a stable (or more stable) nuclide through the process of radioactive decay. A stable nuclide is one that requires the addition of energy to transform it into another nuclide.

Stable nuclides with low mass (apart from hydrogen) have similar number of neutrons and protons but with increasing atomic number there is an increasing excess of neutrons over neutrons and the ‘line of nuclear stability’ deviates from the $N = Z$ line (Figure N.18). The reason for this lies in the characteristics of the nuclear force and the electrostatic repulsion between protons. The nuclear force has a very short range, less than that of a large nucleus, so additional neutrons are required in large nuclei to overcome the electrostatic repulsion between the protons.

**Related Articles:** Nuclear force, Nucleons, Radioactive decay, Stable nuclei

**Nuclear magnetic resonance (NMR)**

*(Magnetic Resonance)* The phenomenon of nuclear magnetic resonance (NMR) was first detected in 1945 independently by Felix Bloch and Edward Purcell (1,2). Both were in 1952 awarded the Nobel prize ‘for their development of new methods for nuclear magnetic precision measurements and discoveries in connection therewith’.

An NMR signal can be detected in odd–odd or odd–even nuclei (i.e. nuclei possessing a magnetic moment or spin) when a sample of the nucleus is exposed to a magnetic field. In the absence of an external magnetic field, all spin states have the same energy but when introduced into a magnetic field $B_0$, a splitting into different possible energy levels occurs, in parallel with a precession of each spin around the $z$-direction (the direction of $B_0$). As an example, the very simple hydrogen nucleus splits into two different energy levels (spin-up and spin-down) and the spins rotate at a frequency of approximately 42.6 MHz/T around $z$. In a sample, a large number of spins contribute to this process and since the rotations of the individual spins are not synchronised (not in phase), a net magnetisation vector ($M_0$) is built up in the $z$-direction (see *Net magnetisation*). Since the rotation frequency (the Larmor frequency) is well defined, it is possible to exchange energy with the system of spins in a sample consisting of a certain kind of nuclei, such as water. If a water sample is exposed to a magnetic field perpendicular to the $B_0$ field (generated by a transmitting coil) and rotating at the exact Larmor frequency (i.e. on resonance), energy is transferred to the sample. The macroscopically observable effect in the sample is that $M_0$ is ‘flipped’ from its original $z$-direction down towards the transverse plane, where it can be detected in a receiver coil using the induction phenomenon. In the early experiments, the frequency of the transmitter was swept over a frequency interval in order to establish the frequency at which the nuclear magnetic resonance phenomenon occurred.

The NMR technique has rapidly evolved into an important tool within chemistry and biochemistry, but has also given rise to the clinically well-established magnetic resonance imaging (see *MRI*) and magnetic resonance spectroscopy techniques. Spectroscopic techniques in clinical as well as non-clinical applications rely on the fact that the Larmor frequency of a specific nucleus depends not only upon the main magnetic field $B_0$ from the magnet, but also upon the immediate atomic and molecular structures surrounding the nucleus since particles such as electrons in the near vicinity of the nucleus have a screening effect. Hence, NMR-detectable nuclei of the same type but situated at different places in a larger molecule show slightly different Larmor frequencies and the corresponding NMR signals can therefore, if frequency resolved (spectroscopically detected), be used to deduce, for example the shape and orientation of the molecule.

**Related Articles:** Magnetic resonance imaging (MRI), Net magnetisation


**Nuclear medicine**

*(Nuclear Medicine)* Nuclear medicine is a special discipline in medicine and medical imaging and it involves the use of specific radioactive isotopes. Nuclear medicine imaging differs from most other imaging modalities like MRI and CT because the images primarily show the physiological function of biological processes rather than morphological structures of the body.

The fundamental idea of nuclear medicine is to target specific biological processes with radiopharmaceuticals (pharmaceutical + radioisotope). Depending on the decay properties of the radioisotope, the radiopharmaceutical can be used for imaging and/or therapy. Radioisotopes with high penetrative radiation are suitable for imaging if the radiation can be acquired with an external imaging detector. For therapeutic purposes, like tumour targeting, radioisotopes with short range are suitable. An increased uptake in an organ or structure might be an indication of pathological condition (see the two examples in Figure N.19).

**Related Articles:** Nuclear medicine

**Figure N.18** Schematic diagram showing the relative position of stable and unstable nuclides. The unstable radionuclides can be classified into three groups, those with excess neutrons, those deficient in neutrons and those with an excess of protons and neutrons.
Newer acquisition techniques allow cardiac gated imaging.

**Abbreviations:** PET = Positron emission gated imaging and SPECT = Single photon emission computed tomography.

**Related Articles:** PET, SPECT, Cardiac gating

**Nuclear reactor**

*(General)* A nuclear reactor is a device in which a nuclear chain reaction is initiated, controlled and sustained. The chain reaction is induced by neutrons and to achieve stability, only one neutron on average, emitted during a fission event, must induce a further fission.

The main components of a nuclear reactor are as follows:

**Fuel:** The most common chain reaction is the induced fission of uranium-235, and the uranium is normally in the form of uranium oxide, encased in tubes to form the fuel rods.

**Moderator:** The cross section for fission of U-235 is low for the fast neutrons emitted during the fission of U-235 but high for slow neutrons. To slow or ‘moderate’ the neutrons, a material which does not capture significant numbers of neutrons is incorporated into the design of the reactor. Water, heavy water, D$_2$O (D = Deuterium), and graphite are suitable and widely used.

**Control rods:** These are made of neutron-absorbing materials, such as boron or cadmium, to control the rate of the chain reaction by insertion or withdrawal.

**Coolant:** A liquid or gas to transfer the heat from the core to outside the reactor. In reactors used for power generation the coolant is used, directly or indirectly, to generate steam to for the turbines generating electricity.

**Core:** It contains the fuel, the moderator, the control rods and the coolant.

**Pressure vessel:** The vessel usually made of steel or concrete, containing the core.

**Shield and containment:** The structure, usually concrete, to contain the pressure vessel and to provide a shield from the intense radiation.

Most reactors are designed as a heat source to produce electricity but other uses include the production of fissile material for making nuclear weapons. For the medical physicist and others involved in the use of radioisotopes in medicine, the main interest in nuclear reactors is their use as an intense source of low- and high-energy neutrons for the production of radionuclides.

The term ‘nuclear reactor’ may also be used for a device in which a nuclear fusion reaction can be initiated, controlled and sustained – a fusion reactor.

**Related Articles:** Cross-section, Nuclear fission, Reactor

**Nuclear Regulatory Commission (NRC)**

*(Radiation Protection)* The Nuclear Regulatory Commission is an independent agency of the USA government.

**Hyperlink:** [www.nrc.gov](http://www.nrc.gov)

**Nuclear transformation**

*(General)* Nuclear transformation is the change of one nuclide into another with a different number of protons or neutrons.

Transformation can be spontaneous, if the nuclide is unstable, through the process of radioactive decay, or induced when a nuclide is bombarded by a particle of sufficient energy or by very high energy gamma rays.

**Related Articles:** Nuclide, Radioactivity

**Nucleon**

*(General)* Nucleon is the collective term applied to protons and neutrons. These are now considered as composite particles, called...
baryons, composed of quarks and gluons but previously were presumed to be elementary particles. Atomic nuclei are composed of protons and neutrons.

**Related Articles:** Baryons, Neutrons, Nuclear forces, Nuclear binding energy, Nucleus, Protons

### Nucleus

(General) The nucleus is the central structure of an atom and is composed of protons and neutrons, which are collectively known as nucleons. The nucleus is surrounded by electrons which circulate around it in orbits, called shells. The nucleus is positively charged due to the protons and is extremely small compared with the overall size of the atom. The nucleus has a diameter on the order of $10^{-14}$ m compared with $10^{-10}$ m for an atom. The nucleus is held together by the nuclear force that is much greater than the electrostatic force repelling the protons.

The simplest nucleus is the nucleus of the hydrogen atom that has one proton and no neutron. Atoms with a small mass have a similar number of protons and neutrons but with increasing atomic number there is an increasing excess of neutrons over protons (Figure N.20).

A simple model of the nucleus is to consider the nucleons arranged in shells, somewhat analogous to the arrangement of the electrons outside the nucleus but only in terms of energy levels. Electron shells are full when they have 2, 8, 18 and 32 in the K, L, M and N shells respectively. In the nucleus the corresponding numbers are 2, 6, 12 and 8; the protons and neutrons have separate shells. When a nucleus has only filled shells an extra nuclear stability is conferred on it, an example is the helium nucleus, with two protons and two neutrons. It is the special stability of the helium nucleus which explains why it is ejected as a particle in alpha decay (Figures N.21 and N.22).

![Simplest representation of the atoms of Hydrogen (1H), Helium (2He) and Lithium (3Li).](image)

Normal nuclei are in the ground state but can, in special situations, exist for a limited time in an excited state. When a nucleus in an excited state falls to the ground state, a gamma ray is emitted. In most cases the lifetime in an excited state is in the range $10^{-6}–10^{-4}$ s but in some situations it can be minutes or hours – in these cases the levels are known as metastable states.

**Related Articles:** Alpha decay, Gamma rays, Protons, Neutrons, Nucleons, Stable nuclei

### Nuclide

(General) The term nuclide is used to denote a species of atom with a specific atomic number $Z$, neutron number $N$ and in a defined nuclear state. Usually it is assumed that the nucleus is in the ground state and the same symbolism is used as for an atom – for example, $^{131}$I represents a nuclide/atom of iodine with total mass (represented by the symbol Z) of 131 and with 53 protons (represented by the symbol N). If the nucleus is in a metastable state then this represented by the superscript $m$, thus $^{131m}$Te is a different nuclide from $^{131}$Te.

**Related Articles:** Atom, Nucleus

### Nude mouse

(Nuclear Medicine) A mouse that lacks a thymus and hence has no mature T cells. Xenografts of human tissue can easily grow in such immune-deficient mice. There also exist nude rats which also are immune-incompetent. These animals are mostly used as experimental animals especially in cancer research in the development of new therapeutic drugs. The nude mouse and rat is hairless, hence the term ‘nude’.

### Number of excitations (NEX)

(Magnetic Resonance) To improve the signal-to-noise ratio (SNR) in magnetic resonance imaging (MRI) it is possible to use two or more excitations/acquisitions of the same k-space line and average the signals together to yield an improved SNR image. The number of used averages is often referred to as the number of excitations (NEX) or the number of acquisitions (NSA).
The SNR in the image will be improved with the square root of the number of excitations; however, the scan time will be prolonged with a factor equal to the NEX value.

**Related Article:** Signal-to-noise ratio (SNR)

### Nyquist frequency

*(Magnetic Resonance)* The Nyquist frequency is the maximum frequency \( \nu_{\text{max}} \) that is correctly encoded by discrete sampling at a frequency \( \nu_{\text{sample}} \):

\[
\nu_{\text{max}} = \frac{\nu_{\text{sample}}}{2}
\]

for real (one channel) signals (as stated by the Nyquist theorem).

Quadrature detection yields also negative frequencies of \( \nu_{\text{min}} = -\frac{\nu_{\text{sample}}}{2} \)

Frequency components exceeding the Nyquist frequency are folded back (aliasing).

**Related Article:** Nyquist theorem

### Nyquist limit

*(Ultrasound)* In a sampling system, the Nyquist limit is always half of the sampling frequency. Frequencies above this limit will not be represented correctly but will appear as a lower frequency.

A frequency which, for example is \( 0.6f_s \), where \( f_s \) is the sampling frequency, will appear as \( 0.4f_s \). The term ‘folding’ is sometimes used, which refers to that the spectrum is ‘folded’ around \( f_s \). In Doppler measurements, ‘folding’, or aliasing, appears somewhat different due to quadrature demodulation, see *Aliasing*.

### Nyquist theorem

*(Magnetic Resonance)* The Nyquist theorem (aka Nyquist–Shannon theorem) states that a band-limited signal of \( \nu < \nu_{\text{max}} \) has to be sampled at a sampling frequency \( \nu_{\text{sample}} > 2\nu_{\text{max}} \).

In other words, digital sampling at a given sampling frequency (or rate) \( \nu_{\text{sample}} \) can correctly encode frequencies below \( \nu_{\text{sample}}/2 \), the so-called Nyquist frequency.

Complex signals acquired by quadrature detection in MRI and MRS allow negative frequencies to be discerned, thus yielding a total bandwidth \( \nu = \nu_{\text{sample}} \times \frac{1}{\tau_{\text{sample}}} \), that is the inverse of the dwell time. Higher or lower frequencies will be folded back below the Nyquist frequency (aliasing).

The Nyquist frequency and theorem are named after Harry Nyquist (1889–1976), Swedish-American engineer.

**Related Article:** Nyquist frequency
Object–film distance

(Diagnostic Radiology) The term object–film distance (OFD) is used in planar x-ray radiology to describe the distance between the central axes of the object being imaged and the surface of the film. The distance influences parameters such as geometric magnification and image distortion. In modern digital radiology where film is no longer used, the distance between the object and the detector is called the object–detector distance (ODD) or the object–image receptor distance (OID) (Figure O.1).

Related Articles: Focal–film distance (FFD), Target–film distance (TFD), Magnification

Object image receptor distance

(Diagnostic Radiology) See Target–film distance

Object recognition

(General) Object recognition is the ability of a computer program to recognize a particular object within an image. In medical imaging, this may be a particular organ such as the liver, lungs, kidneys, etc. This can be achieved by comparing the features of this object (such as shape, size, tracer uptake) with those of a known object.

Object scatter events

(Nuclear Medicine) An object scatter event refers to a photon scattered in the object (i.e. patient or phantom) prior to detection. Consider the situation in image 1, where a photon is emitted with a direction non-parallel to the collimator holes. In an ideal situation, the photon would travel through the object and be attenuated in the collimator. But the photon can undergo Compton scatter and change its propagation direction so that it is parallel to the collimator holes (Figure O.2).

Object scatter events will cause a loss in contrast because of the mis-positioning of events. The mis-positioning can be several centimetres from the original event location. The photon loses energy in the scatter process and it is therefore possible to separate the object scatter events using an energy discrimination window. A scintillation camera with good energy resolution is therefore important.


Oblique

(General) There are a series of terms used to describe the position of an individual when undertaking different imaging examinations. Oblique means turned usually to a specific angle. For example, a 45° angle for an oblique chest projection.

Related Article: Patient position

Oblique imaging

(Magnetic Resonance) Oblique imaging is performed in non-orthogonal planes (or slice orientations) where required by anatomy (e.g. of the foot or heart).

Oblique incidence

(Radiotherapy) Typically, radiotherapy beam data (e.g. depth dose, etc.) is measured with the beam incident at right angles to a flat and uniform surface. However, in reality, for patient treatments this situation is unlikely, and so this data cannot be directly employed to calculate the distribution of dose within the body.

There are two main approaches to compensate for this change: (a) the use of wedges, bolus, or compensators, and (b) a calculation...
technique. Calculation techniques vary from the complex (requiring computer processing) to those which may be calculated manually. Some examples of simple manual calculation correction techniques are the effective SSD method, the TAR method, and the isodose shift method. (For further detail on these correction methods, see Further Readings.) These methods apply for angles of incidence up to 45°.

Abbreviations: SSD = Source-to-skin distance and TAR = Tissue air ratio.

Related Articles: Electron oblique incidence, Incidence angle, Obliquity, Obliquity effect, Bolus, Compensator, Wedge


Obliquity

(Radiotherapy) This is the situation where the patient surface is not perpendicular to the incident beam. The obliquity will therefore cause the dose distribution to be distorted due to the varying distance at which the beam is incident on the skin surface.

Related Articles: Oblique incidence, Electron oblique incidence, Incidence angle, Obliquity effect

Obliquity effect

(Radiotherapy) See Oblique incidence

Related Articles: Oblique incidence, Electron oblique incidence, Incidence angle, Obliquity

Occupancy factor

(Radiation Protection) When designing the shielding for a radiation facility, dose constraints are used. These determine the annual dose in any area outside the facility. Currently, an annual dose constraint of 0.3 mSv is adopted for an area outside the facility.

The dose constraint can be modified (increased) under certain conditions. For example, if the area adjacent to the radiation facility is a controlled area, the dose constraint may be increased. Another permissible modification is to consider occupancy. Some areas of building, e.g. corridors, toilets, etc., would not be expected to have full occupancy. Thus an occupancy factor may be assumed that increases the planning dose constraint. The following occupancy factors are in general use:

- Corridor: 20%
- Toilet (washroom): 10%
- Stairway: 5%

Thus, for example, if the area adjacent to the radiation facility was a corridor, the dose constraint for that area could be increased by a factor of 100/20 (5). Instead of 0.3 mSv, a planning dose constraint of 1.5 mSv could be used.

Related Articles: Dose limits, Dose constraints

Occupational dose limits

(Radiation Protection) The occupational dose limits are set by the International Commission for Radiological Protection (ICRP). The following dose limits apply to exposure from practices, with exception made for exposures from medical practices (diagnostic and therapy) and from natural sources.

Dose Limits for Occupational Exposures: Occupational exposures of any worker shall be controlled in order not to exceed the following limits:

1. Effective dose to whole body: 20 mSv/year averaged over a period of 5 years with a maximum effective dose of 50 mSv in any single year
2. Effective dose to the embryo or foetus: 1 mSv
3. Annual equivalent dose to the lens of eye: 150 mSv
4. Annual equivalent dose in the skin and extremities: 500 mSv

For occupational exposure to radionuclides, with risk of internal contamination, there are tables (for the various radioisotopes) which give ingestion and inhalation dose coefficients. This means that it is possible to evaluate the committed effective dose per unit intake via ingestion corresponding to different gut transfer factors, for various chemical forms, and the committed effective dose per unit intake (via inhalation) for the lung absorption, considering also the component cleared from the lung to the gastrointestinal track.

In the case of apprentices from 16 to 18 years old, who are under training with activity involving exposures, the following limits shall be respected: an effective dose to whole body of 6 mSv/year, an annual equivalent dose to the lens of eye of 50 mSv and an annual equivalent dose to the extremities and skin of 150 mSv.

There might be ‘special circumstances’ under which a temporary change in dose limitation is required and approved. In these cases, the 20 mSv shall be averaged over a period up to 10 years, with the maximum dose per year remaining 50 mSv (circumstances shall be reviewed when any worker reaches 100 mSv). Alternatively, the annual limit shall not exceed 50 mSv with a temporary change period not exceeding 5 years.

Dose Limits for Public Exposures: The averaged estimated doses to critical groups of the public, attributable to practices shall not exceed the following limits: (1) effective dose to the whole body: 1 mSv/year, in special circumstances averaged over a period of 5 years with an effective dose of 5 mSv/year, (2) annual equivalent dose in the lens of the eye of 15 mSv and (3) annual equivalent dose to the skin of 50 mSv.

Dose Limits for Comforters and Visitors of Patients: The dose of any voluntary comforter or visitor of patients shall be constrained so that it is unlikely to exceed 5 mSv during the period of diagnostic investigation or treatment. The dose to children visiting patients who have ingested radioactive material shall be limited to less than 1 mSv.

Related Articles: International Commission for Radiological Protection (ICRP), Occupational exposure


Occupational exposures

(Radiation Protection) Exposures of workers incurred in the course of their work are considered occupational exposures.

Responsibilities: Different main responsibilities are attributed to registrants, licensees and employees. Licensees shall ensure that
for all workers (1) occupational exposures are limited and optimised; (2) suitable and adequate facilities, equipment and services for protection are provided; (3) appropriate protective devices and monitoring equipment are provided and properly used; (4) appropriate training is provided as well as periodic retraining and updating; (5) adequate records are maintained.

Employees shall follow any applicable rules for protection, use the monitoring devices, protective equipment and clothing provided properly and co-operate with the licensee with respect to protection.

Conditions of Service: Special compensatory arrangements (preferential treatment, salary, vacation, etc.) shall be neither granted nor used as substitutes for the provision of proper protection and safety. The condition of work of pregnant women should be adapted in order to ensure the compliance with the dose limit for the embryo/foetus.

No person under age 16 years shall be subject to occupational exposures, and no person under the age of 18 years shall be allowed to work in a controlled area unless supervised (under training). Suitable alternative employment should be provided to workers who may no longer continue in employment involving occupational exposure.

See Dose limits for the specification of the values.


Hyperlink: http://www.IAEA.org

Oersted
(General) Oersted (Oe) is a cgs unit for magnetic field strength. 1 Oe = 1000/4π AT/m (cgs – the centimetre, gram second system of units, the cgs is now replaced by SI).

Oe is related to the cgs unit gauss (G) through the magnetic permeability \( \mu \), (relative to permeability of vacuum):

\[
B (G) = \mu, * H (Oe)
\]

The Oersted is named after the Danish physicist Hans Christian Oersted.

Related Article: Gauss

OFD (object–film distance)
(Diagnostic Radiology) See Object–film distance

Off-resonance
(Magnetic Resonance) Off-resonance occurs when spin isochromats have a frequency different from that expected, i.e. the Larmor precessional frequency. For example, the Larmor frequency of protons in fat differs from that of protons in water due to different magnetic field strengths. However, even if the Larmor frequencies of all protons were equal, differences in magnetic susceptibility throughout the object would result in local magnetic field gradients within the object, leading to differences in the Larmor frequency. Other sources of off-resonance effects can be eddy currents arising in the conducting material of the scanner when gradients are switched on and off, or concomitant gradients (Maxwell terms), e.g. a magnetic field gradient applied in the z-direction results in additional gradients in the x- and y-direction as well since all magnetic fields must obey Maxwell’s equation.

The effects of off-resonance include spatial distortions (e.g. chemical shift artefacts or artefacts occurring in the surrounding of metal implants), signal losses (c.f. \( T_2^* \) relaxation) and blurring, depending on the k-space trajectory. In general, spin echoes reduce off-resonance effects in comparison to gradient echoes. In ultrafast imaging (where EPI-readouts are particularly sensitive), reducing the readout time will reduce these effects. Other sequences vulnerable to off-resonance effects are so-called balanced gradient echo sequences and spectroscopic sequences, both requiring adequate shimming for achievement of optimal results.

Related Article: Eddy currents

Offset
(General) In electronics, offset means offsetting of a signal from zero. It is considered to be direct current voltage measured at the input or output of an electronic device. In operational amplifiers, input offset voltage and input offset current are specified for each type.

Direct current (DC) offset is usually undesirable, but can be eliminated by trimming or by auto-zero amplifiers in electronic devices.

On-line portal imaging
(Radiotherapy) On-line portal imaging involves the use of a portal imaging device to make measurements of treatment accuracy, in terms of patient positioning, during a fraction of radiotherapy and to act on that information to reduce errors for that treatment fraction, hence the name ‘on-line’. Often this is achieved by delivering a small test dose at the start of the fraction, imaging it, determining the set-up correction needed and making that correction before delivering the rest of the treatment.

On-Line versus Off-Line Set-Up Correction: The alternative to on-line imaging is to image a treatment fraction and determine the set-up error off-line, after the fraction has finished, and correct subsequent fractions on this basis. In practice several fractions are imaged before any corrective action is taken for the off-line process so that random errors in each image are averaged out and the systematic error is determined and corrected. Various empirically derived methods exist for deciding how many images to take and how often to take those images to arrive at the most accurate treatment with the minimum workload for off-line set-up correction.

Errors and Margins: The overall error in a treatment has many contributing sources. The set-up errors detected with on-line portal imaging arise from several of these sources. Radiotherapy treatment errors may be described in terms of random and systematic components. Crudely, the random component detected in portal imaging is the day-to-day variation in set-up and the systematic component is the difference between the average position at treatment and that at planning. On-line correction addresses both types of errors, whereas off-line correction addresses only systematic errors.

Various numerical simulations have been carried out to determine the treatment margins necessary to achieve a certain probability of target coverage as a function of the size of the two types of errors. These reveal the effects of systematic errors to be three times more important than random errors.
The one-day low-dose stress/high-dose rest sequence, on the other hand, has the advantage of requiring image acquisition durations similar to those used for $^{201}$Tl, making it easy for a laboratory to alternate between stress/distribution $^{201}$Tl and stress/rest $^{99m}$Tc-Sestamibi or $^{99m}$Tc-Tetrofosmin. This sequence is also a good choice for patients with low likelihood of coronary artery disease, since the second study is not always needed if the stress study is normal. The disadvantage of this sequence, on the other hand, is that the count rates in the stress image set are not adequate for accurate assessment of defect reversibility. If the stress/rest sequence is chosen, a longer interval between the two injections is required so that the background contribution from the first set of images is minimised.

**Related Article:** Two-day protocol


**One-way rectifier**

(Diagnostic Radiology) One-way rectification (also known as half-way rectification) is an electric power supply which uses either the positive or the negative AC wave. Half-way rectifiers use either one diode (single-phase power supply) or three diodes (three-phase power supply). This type of rectification is relatively inefficient, because it blocks half of the input signal (Figure O.3).

**Related Article:** Rectifier

**Opacity**

(Diagnostic Radiology) Opacity is the characteristic of an object or substance to reduce the penetration or passage of radiation such as light. It is the inverse of transparency. The degree of opacity can be measured and expressed in terms of the optical density. Optical density is used extensively to express the ‘opacity’ of areas within a radiographic film.

**Open field**

(Radiotherapy) Open field is the term generally applied to a field used with no beam modifier present in the beam, such as wedge, compensator, tray or shielding blocks. The open field obtained with secondary collimator or multileaf collimator may be square, rectangular or irregular.

**Related Articles:** Secondary collimator, Multileaf collimator, Block design, Wedge, Compensator, Irregular field

Open-core transformer

(Diagnostic Radiology) See Transformer

Operating mode

(Diagnostic Radiology) See Acquisition modes for digital image

Operational amplifier

(General) An operational amplifier is high-gain DC-coupled electronic voltage amplifier. It amplifies the voltage difference between its differential inputs (+ input and − input). It has a single voltage output. Operational amplifiers have very high input impedance at its differential inputs (+ input and − input). It has a single voltage output terminals and very low output impedance (Figure O.4).

Operational amplifiers have very high input impedance at its differential inputs (+ input and − input). It has a single voltage output. Operational amplifiers have very high input impedance at its differential inputs (+ input and − input). It has a single voltage output terminals and very low output impedance (Figure O.4).

Operational amplifiers are used in a number of different configurations, but primarily as amplifiers. The voltage gain of an operational amplifier is determined by negative feedback. Typical configurations of DC voltage amplifiers are shown in Figures O.5 and O.6.

The gain of the non-inverting amplifier is determined by $G = 1 + \frac{R_2}{R_1}$ and the gain of the inverting amplifier by $G = -\frac{R_2}{R_1}$.

In Figure O.7, the differential amplifier configuration is presented. Differential amplifiers are often used in sensing circuits to amplify low-level voltage difference signals whilst rejecting any common voltage.

Operational amplifiers are most widely used analog integrated circuits (IC) with applications in industrial, professional, consumer and medical electronic devices. They are encapsulated in many standard IC housings, typically in 8-pin dual in-line packages.

Optical density

(Diagnostic Radiology) Optical density is the opaqueness of translucent film and has assigned numerical values related to the amount of light that penetrates the film. Increasing film density decreases light penetration. The relationship between density values and light penetration is exponential, as shown in Figure O.8.

A clear piece of film that allows 100% of the light to penetrate has a density value of 0. Radiographic film is never completely clear. The minimum film density is usually in the range of 0.1–0.2 density units. This is designated 'the base plus fog density' and is the density of the film base and any inherent fog not associated with exposure.

Each unit of density decreases light penetration by a factor of 10. A film area with a density value of 1 allows 10% of the light to penetrate and generally appears as a medium grey when placed on a conventional viewbox.

A film area with a density value of 2 allows 10% of 10% (1.0%) light penetration and appears as a relatively dark area when viewed in the usual manner. With normal viewbox illumination, it is possible to see through areas of film with density values of up to approximately 2 units.

Optical densities of 3 (or even the maximum 4) cannot be distinguished visually. They all appear completely black, but can be measured with optical densitometer.

Optical distance indicator

(Radiotherapy) The optical distance indicator (ODI) allows the operator to know the distance from the source of radiation to the skin (or phantom) surface. It is commonly a scale pattern projected by a light source in the head of the linac, and is tilted at an angle such that the appropriate point on the scale is in focus on the central axis at the corresponding SSD. Therefore, at the isocentre (SSD = 100 cm) the 100 indicator is focussed on the central axis, while at 10 cm above and below the isocentre the 90 and 110 indicators, respectively, are focussed on the central axis. The scale typically runs from around 80 to 150 cm, to allow for a range of different SSD set-ups, e.g. isocentric treatments and extended SSD treatments where fields of size greater than 40 cm are needed. The level of agreement of the ODI with SSD (and couch vertical movement) is usually of the order of 3 mm towards the extremes of the range and 1 mm at the isocentre (100 cm).

Related Article: Source-to-skin distance (SSD)

Optical transfer function

(Diagnostic Radiology) The optical transfer function is the Fourier transform of the (real) point-spread function. Its magnitude is the
Light source, e.g. laser or LED

Previously irradiated by ionizing radiation crystal, e.g. quartz

Optical filter

Electronic processing of data

PC (output presentation)

FIGURE O.9 A scheme of a set-up for OSL radiation dosimetry.

Optically stimulated luminescence

is a physical phenomenon of the emission of optical radiation in the form of prompt fluorescence, stimulated by illuminating a previously irradiated sample. Many crystals, e.g. quartz, aluminium oxide and feldspars, can absorb energy of the ionising radiation and store it by elevating electrons from the valence to the conduction band, then capturing it at trapping centres (metastable states) within the bandgap, below the bottom of the conduction band. Holes, created in the valence band, can move through the crystal and be trapped by trapping centres situated above the top of the valence band. The lifetime of the electron or hole traps can be large (up to hundreds of years). If the trap is a luminescence centre, then the energy of the optical radiation from a laser or an LED (light-emitting diode) will cause de-excitation (recombination) and, consequently, emission of light (luminescence) will occur. The emitted light is detected using a photomultiplier tube. The equipment needed in OSL dosimetry consists of a readout light source (laser or LED) for exciting charges from traps and initiating recombination, a set of optical filters to shield the photomultiplier tube from the direct or reflected light from the readout source, a photomultiplier tube measuring the light emitted by the previously irradiated sample, associated electronics and a PC (Figure O.9).

The readout light source can be used as a continuous excitation source or as multiple pulses of light. The readout time is of the order of several seconds.

The sensitivity level is down to a few μGy for OSL radiation dosimetry using aluminium oxide.

The OSL technique can also be applied to optical dating of ancient materials.

Related Articles: Dosemeter, Dosimeter, Thermoluminescent dosimeter

Optimisation

(Radiation Protection) The second principle of protection against ionising radiation for workers, patients and members of the public, specified by the International Commission on Radiological Protection, is optimisation.

Once a practice, carried out by an employer involving the exposure of staff, patients or the public, has been justified in terms of the net benefit outweighing the risks associated with the exposure (Justification), then it must be optimised – i.e. the exposure must be minimised whilst the benefit of the practice is still achieved. This is often called the ALARP Principle (that doses should be as low as reasonably practicable). It used to be called the ALARA Principle (as low as reasonably achievable). The radiation employer is entitled to consider economic and social factors when deciding what measures to implement to try to reduce doses to staff, patients or the public – hence the use of the word ‘practicable’.

Finally, once a practice has been justified and optimised, the employer must also ensure that any radiation doses received by staff or the public are under statutory dose limits – i.e. limitation.

Related Articles: ALARP principle, Justification, Limitation


Optocoupler

(General) An optocoupler is an electronic device used to electrically isolate two parts of a system, while allowing data transmission from one side to the other. Optocouplers are available in integrated circuit packages as shown below. An LED transmitter on the input side converts input signals to light and transmits this light across a transparent barrier to a phototransistor receiver. The phototransistor senses this light and an output signal can be generated in external circuitry. Optocouplers can be configured to transmit either analog or digital data from input to output.

The barrier between the transmitter and receiver in an optocoupler is non-conducting and electrically separates the two sides.

Optocouplers find application in ECG monitoring to electrically isolate the ECG amplifier connected to the patient from the rest of the device. This safety measure eliminates any potential electrical paths from the machine to the patient leads (Figure O.11).

Ordered subset expectation maximum method

(Nuclear Medicine) The main problem with the maximum likelihood expectation maximisation (MLEM) method is the long time required to obtain acceptable accuracy in the reconstructed image. A successful method to accelerate the convergence rate is the ordered subset expectation maximum (OSEM) method. This method is identical to the MLEM algorithm in its principles but differs at the stage where the image is updated. In the MLEM algorithm, the image is updated (i.e. multiplied by the correction matrix) only after all projection angles have been processed.

In the OSEM algorithm, the image is instead updated after a subset of projections has been processed. For example, a common number of subsets are 16 for a 64 projection angle acquisition. Hence, the image is updated after four angles have been processed. The acceleration in this method is roughly proportional to the number of subsets. The order of the projection angle is not linear but follows a particular pattern for an optimal reconstruction. An iteration is defined when all subsets are processed. The process is then repeated for the desired numbers of iterations. Figure O.12 shows an example using images obtained with different iteration numbers for MLEM and OSEM.

Related Article: Maximum likelihood expectation maximum (MLEM)
Organ at risk (OAR) (Radiotherapy) Any normal tissue whose radiosensitivity significantly influences the treatment planning or prescribed dose is known as an organ at risk (OAR). As for the target volume, it is important to consider the physiological movement and set up errors and incorporate this variability within a margin.

The use of OARs as planning volumes was proposed by the ICRU in Report 50 (with addendum 62). This report provided a common framework on prescribing, recording and reporting therapies, with the aim to improve the consistency and inter-site comparability. This report details the minimum set of data required to be able to adequately assess treatments without having to return to the original centre for extra information.

OARs can be classified into two categories – parallel or serial organs:

1. Parallel organs can be treated using a ‘critical volume’ model; the organ architecture is such that each individual element is not crucial to the overall function of the organ. It is important to ensure that a significant volume of the organ is not damaged by the radiation. Tolerances may vary with percentage volume irradiated. Examples include the liver, kidney and lung.

2. In contrast, the spinal cord and rectum are serial organs, following a ‘critical element’ model, each element is crucial to the function. If one element is damaged then the organ is damaged irreparably.

Abbreviation: OAR = Organ at risk.


Organ dose (Radiation Protection) Organ dose is otherwise known as the equivalent dose, H, and is a product of the absorbed dose D to an organ and the relative biological effectiveness (RBE) for the irradiation:

\[ H = D \times \text{RBE} \]

For radiation protection purposes, the RBE is taken into account by the use of the radiation weighting factor. Since RBE = 1 for the vast majority of diagnostic and therapeutic radiations, organ dose is usually equivalent to the absorbed dose to the organ.

Risks from ionising radiation are based on calculations of organ doses. In therapeutic applications, the organ dose to critical tissues (peripheral tissue of high radiosensitivity, e.g. bone marrow) is the limiting factor, and the efficacy of the treatment is usually determined by this.

Methods of estimating organ doses vary depending on the purpose of irradiation. Very accurate dose estimates are made in radiotherapy where large doses prescribed to tumours must be carefully optimised to minimise surrounding critical-organ doses. In diagnostic procedures, where the accuracy is less critical, organ doses are usually estimated by using the scan parameters in a calculation that assumes a standard-sized patient phantom.

Related Articles: Radiosensitivity, Relative biological effectiveness, Radiation weighting factors

Organic liquid scintillators (Radiation Protection) Liquid scintillation counting is used for detection of a weak penetrating radiation like beta or alpha particles.

There are several types of organic scintillators, e.g. crystals, liquid organic solution, plastic, thin plastic films and ones loaded with high atomic number elements, such as Pb or Sn. Organic scintillators are transparent to their own fluorescence emitted radiation.

Liquid organic scintillators consist of an organic scintillator, e.g. anthracene, solvent and sometimes a wavelength shifter if it is necessary to adjust the light wavelength to the spectral sensitivity of the photomultiplier. They can be used up to very high exposures, e.g. 10^7 Gy.

The counting efficiency, i.e. the scintillation efficiency, is defined as a ratio of the number of particles whose energy is converted in the light to the number of all incident particles. If the radioactive material, e.g. C-14 or H-3, is dissolved in a liquid scintillator, the counting efficiency attains 100%.

Liquid organic scintillators with a hydrogen-containing organic solvent are applied in the detection of fast neutrons using proton recoil.

Related Article: Liquid scintillation (LS) counting


Ortho pan tomography (OPG) (Diagnostic Radiology) Ortho pan tomography, or orthopantomography (OPG), is a dental x-ray procedure that scans the curved jaws and teeth and produces a panorama-type image on a flat film or digital receptor.

An OPG system is shown in Figure O.13. OPG equipment includes x-ray tube with slit beam restrictor, which produces thin vertical beam with width of the order of 4–7 mm (at the film/detector surface) and height sufficient to cover the mandibles and maxilla regions. The x-ray tube rotates around the patient head and the x-ray beam exposes the x-ray film/detector positioned opposite to the x-ray tube. In front of the x-ray film/detector is another slit diaphragm, which allows only a thin x-ray beam to expose part of the film/detector. During the rotation of the x-ray tube around the patient head, the film/detector moves behind this slit diaphragm; this way, different parts of the film/detector record the x-ray beam passing through different areas of the patient head. The produced panoramic radiograph of the jaws and teeth (Figure O.14) usually

![FIGURE O.13 A typical OPG x-ray system.](image-url)
delivers high dose over the scanned area, as the rotation around the head (the x-ray exposure) usually takes several seconds.

**Orthogonal films**

*Radiotherapy, Brachytherapy*

**Definition of Applicator and Source Positions:** Radiographs are commonly used for reconstruction of applicator/source positions when image-guided brachytherapy techniques are not available. At least two radiographs are needed for the 3D reconstruction. Orthogonal radiographs, frontal (anterior to posterior, AP) and lateral, are the most commonly used imaging arrangement.

Two orthogonal radiographs are shown in the article entitled *Internal reference point*. These are used to determine rectum and bladder reference points for a cervix cancer application.

Reconstruction devices, such as the jig shown in Figure O.15a, have also been implemented in some brachytherapy treatment planning systems. A reconstruction jig, with its markers, has a stable...

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**FIGURE O.14** Panoramic radiograph of the jaws and teeth (OPG radiograph).

**FIGURE O.15** (a) Lateral radiograph (one of two orthogonal films) with marker wires inserted in the two treatment channels. The spinal cord is indicated by a thin pencil line by the oncologist. The reconstruction jig to be used is shown to the right, a stable 90° jig arm with two boxes, each with two crosses.

(b) Lateral and frontal radiographs with the jig in place showing the two crosses on each box. Marker wires are inserted in the two treatment channels. Lead shot marks the medial and lateral corners of the right eye. Also shown are the isocentre cross and square belonging to the x-ray unit.
Orthovoltage radiation
(Radiotherapy) Orthovoltage therapy describes treatment with medium energy kilovoltage x-ray beams (accelerating potential 160–300 kV; 0.5–0.4 mm Cu half-value layer [HVL]).

There is no sharp division between the various voltage ranges and they can vary slightly in different documents.

Abbreviation: HVL=Half-value layer.


Oscillating gradient
(Magnetic Resonance) An oscillating gradient is used to fill k-space using a spiral trajectory. The spiral EPI sequence uses two simultaneously applied magnetic field gradients, and the data are collected continuously during an acquisition time $T_D$.

The method offers reduced sensitivity to motion and some reduction in imaging time, and does not require such strong gradients (Figure O.16).


Related Articles: $k$-space, Spiral scanning

Oscillator
(General) An electronic circuit that converts DC power into AC power at a frequency defined by values of electronic components used in the circuit. Basic types of oscillators are harmonic and relaxation oscillators. Harmonic oscillators generate sinusoidal waveforms, while relaxation oscillators generate non-sinusoidal waveforms such as square wave, sawtooth, etc.

The block diagram of an oscillator in Figure O.17 comprises of an amplifier having gain $A$ with external positive feedback circuit $\beta$.

Related Article: Operational amplifier

Oscilloscope
(General) Oscilloscope is an instrument that enables the display and measurement of signal parameters from a screen. Most oscilloscopes have linear time bases that enable measurement of the parameters against time, but many enable measurements with respect to other electrical quantities as well. Analog oscilloscopes display the waveforms on the screen of a cathode ray tube. Digital oscilloscopes retain the basic idea of viewing the signal waveform on the screen, but use a microprocessor or DSP computational power to calculate a vast number of signal parameters. Digital oscilloscopes usually can store the waveform and the measurement results in a memory. Their input signal range is regularly from a few millivolts to a few tens of volts, and the frequency range from DC to gigahertz (Figures O.19 and O.20).

Osmosis
(Nuclear Medicine) The diffusion of water molecules through a semipermeable barrier, e.g. a cell membrane, is called osmosis. One prerequisite for osmosis is two compartments with different solute concentrations leading to a flow of water molecules from one compartment to the other. Osmosis can also be defined as the net flow of water molecules from a compartment with high water potential (i.e. a measurement of the tendency of water to move between compartments) through a semipermeable barrier to a compartment with low water potential.

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**FIGURE O.16** Sequence diagram for a spiral EPI sequence.

**FIGURE O.17** Block diagram of an oscillator.

**FIGURE O.18** Electrical schematic of Wien-bridge oscillator built with an operational amplifier.

In order to maintain steady oscillations, the Barkhausen equation must be satisfied:

$$A \cdot \beta = 1 \angle 0^\circ$$

Feedback oscillators are often realised with operational amplifiers. $RC$ circuits ($C_2R_4$ and $C_1R_3$) form the positive feedback network and the gain is determined by the resistor network $R_1$ and $R_2$ (Figure O.18).

Related Article: Operational amplifier

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Output factor

(Radiotherapy) Linear accelerators are calibrated to deliver a known dose at a specific point in a given field (under reference conditions). For example, it is common to calibrate electron beams at the reference depth \( d_{ref} \) for a \( 10 \text{cm} \times 10 \text{cm} \) field at SSD = 100 cm. All other dose data, as used in dose calculations, can be then calculated from a ratio of the current output to the output at the reference depth under reference conditions. This ratio is known as the output factor, and it is used to calculate the length of time of beam-on, or monitor units (MUs), that must be used to deliver the prescription dose.

For example, if the output factor at the prescription point is 0.5, then the number of MUs needs to be doubled in order to deliver equal dose to the PTV, as it would have done at the reference depth under reference conditions.

The output factors are usually defined as the ratio of the dose for non-reference conditions to the dose under reference conditions. They may be determined as the ratio of corrected dosimeter readings measured under a given set of non-reference conditions to that measured under reference conditions. The measurements are usually done at the depth of maximum dose or at the reference depth and corrected to the depth of maximum dose using PDD (or TMR). This method minimises the dosimetric error due to positioning and reduces any electron contamination effect. Reference conditions depend on the type of ionising radiation (photon beams, electron beams, kilovoltage x-ray beams, etc.) and on the implemented concept of the absorbed dose determination. The output factors are measured and tabulated at commissioning. They take into account the field size and the field shape, wedges, off-axis positioning, various SSD, etc.

**Abbreviations:** MU = Monitor unit, PDD = Percentage depth dose, PTV = Planning target volume, SSD = Source surface distance and TMR = Tissue maximum ratio.

**Related Articles:** Percentage depth dose, Electron contamination

Output screen

(Diagnostic Radiology) See Image intensifier

Over-table radiography

(Diagnostic Radiology) Over-table (over-couch or overtable) radiography uses x-ray tube mounted above the patient. This is the most typical setup of x-ray radiographic equipment. However, for fluoroscopic equipment, this set-up produces considerably more scattered radiation (from the patient) to the staff, and in this case, under-couch x-ray tube positioning is the preferred option. Some sources also name these mounts according to the detector (e.g. under-couch film holder, or over-couch image intensifier).

**Related Articles:** Radiography, Fluoroscopy

Over-table tube

(Diagnostic Radiology) See Over-table radiography

Over voltage

(Diagnostic Radiology) See Overvoltage protection

Overbeaming

(Diagnostic Radiology) In CT scanning, the term ‘overbeaming’ is used to refer to the additional extent of the z-axis x-ray beam profile over the nominal z-axis beam profile, as shown in Figure O.21.

Overbeaming usually occurs on multislice CT scanners and is due to the x-ray beam penumbra. In multislice CT, all detector banks used for image acquisition should receive a uniform photon flux if a uniform image quality is to be achieved. The x-ray beam extent is therefore increased so that the penumbra lies outside the active detectors, shown in dark grey in Figure O.21. The extent of penumbra on CT scanners is typically in the order of 2–3 mm.

Overbeaming results in a reduction in dose efficiency as not all the photons that the patient is exposed to contribute to the image.

**Related Articles:** Geometric efficiency, Multislice CT scanner

Overexposed

(Diagnostic Radiology) See Overexposure

Overexposure

(Diagnostic Radiology) Overexposure is a condition in which something has received more than the optimum radiation exposure. For most medical procedures using radiation, there is usually an optimum amount of exposure that produces the necessary image quality. In film radiography, overexposure results in excessive film density or darkness and a possible reduction in contrast. In digital radiography overexposure of the receptor (above the optimum value) results in unnecessary exposure to the patient.

Radiographs produced with three different exposures are compared in Figure O.22.
Overload

*Diagnostic Radiology*  See Overload protection

**Overload protection**  
*Diagnostic Radiology*  Overload is the excess of energy imparted to the anode of an x-ray tube. Each tube has its maximal tube load parameters (kV, mA and overall watts or heat units). These are specific for each focal spot. Loading over of the thermal capacity of the specific actual focal spot can lead to the local melting of anode. This can further lead to destruction of the whole x-ray tube. Each x-ray tube has a special circuit for overload protection.

There are different ways to assure the protection of the x-ray tube anode against overloading. One of the simplest systems is a rubber membrane in the x-ray housing allowing expansion of the insulating/cooling oil (due to excessive heat) to activate a cut-off switch.

However, the protection has to be related to the specific actual focal spot. In this case, a calculator estimates the maximal kV and mA s, compares these with the x-ray tube rating chart and in case of potential overloading does not allow for the exposure. Similarly, different charts will be used for overload protection in multiple exposures (e.g. in angiographic series).

Older x-ray equipment use special capacitor whose charge represents the heat units of the anode. The capacitor charges during the exposure and after it discharges through a system with time constant similar to the cooling curve of the x-ray tube. In this way, a system measuring the charge of the capacitor will effectively show the accumulated heat in the anode. This system can be used to disallow (or interrupt) exposures which could overload the x-ray tube. In a similar way, contemporary x-ray equipment use micro-processor simulator, which works in parallel with the x-ray tube and shows its current thermal load. The simulator provides signal for the overload protection system. Normally, replacing of the x-ray tube requires replacement of memory (or firmware) storing information about the thermal capacity of the tube.

**Related Articles:**  Heat units, Heat capacity, Maximum load, Cooling curve, Tube rating charts, Actual focal spot, Overload

**Oversampling**  
*Magnetic Resonance*  Oversampling denotes sampling of a band-limited signal at a rate faster than required by the Nyquist theorem. Since aliasing of unwanted components exceeding the Nyquist frequency is avoided, these signals can be removed after Fourier transform. In MRI, oversampling (usually by a factor of 2) is routinely performed during readout. In this direction, the field of view may be chosen to be smaller than the object. In 3D MRI, oversampling may be performed in the slice-direction to avoid aliasing.
Overscanning

Related Articles: Nyquist frequency, Fourier transform

Overscanning

(Diagnostic Radiology) In CT, ‘overscanning’ refers to the additional irradiation in helical scans from x-ray tube rotations beyond the extent of the nominal selected scan length (Figure O.23).

Some overscanning is required by necessity in a helical scan, so as to obtain sufficient data for interpolation of axial images from the helical data set (see Spiral interpolation).

The increase in dose due to overscanning is more significant on multislice scanners due to the greater extent of the z-axis x-ray beam profile (Figure O.23b). It is also more significant for short scan lengths where it may be more dose efficient to employ a narrower z-axis x-ray beam despite the reduced dose efficiency of narrow collimations (see Geometric efficiency). Alternatively, where appropriate, sequential (axial) scanning may be employed which does not require the additional irradiation.

Recently, some manufacturers have adopted a method of ‘adaptive collimation’ to minimise the extent of the additional irradiation from overscanning.

Related Articles: Spiral interpolation, Geometric efficiency

Overshoot

(General) Overshoot is a characteristic of a realistic rectangular electrical pulse, when the instantaneous value of the pulse rises to a value larger than amplitude (height) of the pulse and then decays to the amplitude value usually in a damped oscillatory way (Figure O.24).

Related Article: Pulse

Oxgen enhancement ratio

(Radiotherapy) The oxygen enhancement ratio (OER) is the ratio of the radiation dose given under hypoxic conditions to the radiation dose given under aerobic conditions.

![FIGURE O.23](See colour insert.) Overscanning due to additional irradiation beyond the nominal scan length (a) on a single slice scanner and (b) on a multislice scanner. (Graphs courtesy of ImPACT, UK, www.impactscan.org)

![FIGURE O.24](Overshoot of a rectangular pulse.)

Oxygen

(General)

Symbol: O
Element category: Non-metals
Mass number A: 16
Atomic number Z: 8
Atomic weight: 15.994 g/mol
Electronic configuration: 1s2 2s2 2p4
Melting point: 54.8 K
Boiling point: 90.2 K
Density near room temperature: 1.43 kg/m³

At standard temperature and pressure, oxygen is a diatomic gas, with molecular formula O₂, which is both colourless and odourless. Oxygen is used by all living cells in the process of respiration, combining glucose and oxygen to produce carbon dioxide, water and stored energy in the form of adenosine triphosphate (ATP).

Isotopes of Oxygen and Their Medical Applications:

Oxygen has three stable isotopes all of which have medical applications. Radioactive ¹³N, used in positron emission tomography (PET), is produced using ¹⁶O. ¹⁷O can be used as a tracer to study cerebral oxygen utilisation. ¹⁸O is used in cyclotrons to produce ¹⁸F which can be incorporated into fluorodeoxyglucose (FDG), a common PET radiopharmaceutical. In addition, cyclotron produced ¹⁸O can be used in the form of ¹⁸O-Water to assess myocardial perfusion.

In functional MRI, the relative change in blood oxygenation level in different areas of the brain during mental tasks is utilised as a form of contrast sensitive to brain activation.

Related Articles: Blood oxygenation level–dependent contrast, Cyclotron, Functional magnetic resonance imaging, Isotope, Myocardial perfusion imaging, Oxygen enhancement ratio, Oxyhaemoglobin, Positron emission tomography, Reoxygenation, Radioactivity

Oxygen enhancement ratio

The oxygen enhancement ratio (OER) is the ratio of the radiation dose given under hypoxic conditions to the radiation dose given under aerobic conditions.
dose given under fully oxygenated conditions that achieves the same biological effect.

It is known that tumour cells can be hypoxic and therefore less sensitive to radiation causing DNA damage indirectly through free radicals produced by the ionisation of oxygen. Therefore, the presence or absence of oxygen dramatically influences the biological effect of sparsely ionising radiations (low-LET) such as x-rays but there is no effect for densely ionising radiations (high-LET) such as α-particles. With low-LET radiation, the addition of oxygen increases radiosensitivity, and the precise value of this factor is the OER. For low-LET radiation the OER is 2.5–3.5 and for high-LET radiation it is unity. For intermediate LET radiation, such as neutrons, it is about 1.6.

Studies of the effects of oxygen on the response of rapidly growing cells cultured in vitro to x-rays suggest that the OER has a value of 2.5 at lower doses such as those typical of the daily dose per fraction delivered in fractionated radiotherapy regimes (around 2Gy). This is believed to be due to a variation of OER with the phase of the cell cycle. The OER is lower for cells in the G2 phase (value about 2.3) than in those in the S phase (value about 2.8). The G2 cells are more radiosensitive and therefore dominate the low-dose region of the survival curve. Therefore, the OER of an asynchronous cell population is slightly smaller at low doses than high doses. This has been demonstrated in vitro where precise survival measurements are possible but would be difficult to show in tissue. Although this is an interesting radiobiological observation, it is of little or no clinical significance in radiotherapy.

Abbreviations: LET = Linear energy transfer and OER = Oxygen enhancement ratio.

Related Articles: Cell cycle, Linear energy transfer (LET), Linear quadratic (LQ) model, Radiosensitivity, Redistribution, Reoxygenation


Oxyhaemoglobin

(Magnetic Resonance) Haemoglobin is an oxygen-transporting protein attached to red blood cells. The essential active component of haemoglobin is iron, a transition element with variable electronic configuration. As such, the iron within haemoglobin allows the transformation of the protein from an oxygen-rich diamagnetic substance (oxyhaemoglobin) to an oxygen-poor paramagnetic substance (deoxygenated haemoglobin). In the diamagnetic substance there are no net unpaired electrons (the electrons pair in spin-up/spin-down partnerships according to the Pauli’s exclusion principle). The paramagnetic substance has the six outer electrons of the iron ion distributed across the five 3d orbitals in the form of one pair and four single electrons of identical spin. It is these four unpaired electrons of identical spin which cause the deoxyhaemoglobin to be paramagnetic. When the haemoglobin molecule is oxygenated, there is a slight structural change which alters the lowest energy electronic configuration: the electrons then have no net spin-up/spin-down imbalance.

In many areas of the body, a healthy reserve of oxyhaemoglobin can be stored in muscle, but in the brain, there is insufficient muscle for this to be the case: storage is poor. Thus, to protect the brain’s supply of oxygen, oxyhaemoglobin is always oversupplied (even in periods of rest).

During periods of neuronal stimulation, the oversupply is escalated so that even after the brain has taken the oxygen it needs, the venous blood flow is still more oxygenated during activity than during the resting state.

As oxyhaemoglobin has no net electron spin imbalance, it dephases more slowly than deoxyhaemoglobin. This means that increased levels of oxyhaemoglobin cause increased $T_1$ and $T_2^*$ at sites of neuronal activity. fMRI pulse sequences are usually $T_2^*$-weighted EPI, such that increased oxyhaemoglobin leads to increased MR signal.

Related Articles: fMRI (functional magnetic resonance imaging), Blood oxygenation level–dependent contrast, BOLD, Haemodynamic response function
P-32-sodium orthophosphate
(Nuclear Medicine) Phosphorus-32 is a reactor produced radionuclide through the reaction $^{31}\text{P}(n, \gamma)^{32}\text{P} \xrightarrow{14.26\text{d}} ^{32}\text{S}$. It is a $\beta$-emitter with maximum beta energy of 1.711 MeV. The half-life of $^{32}$P is 14.26 days. It is obtained as a solution of $^{32}$P-sodium phosphate suitable for oral or intravenous administration. The solution is clear and colourless with pH 5–6. It contains isotonic saline and a sodium acetate buffer. In nuclear medicine $^{32}$P is mainly used for radionuclide therapy of polycythemia vera (a rare blood disease characterised by an elevation of the immature red blood cells) and other neoplastic haematologic diseases. It is frequently in use in biomed-ical research as a tracer substitute for phosphor, e.g. in nucleic acid sequences.

The $^{32}$P-sodium phosphate activity recommended for therapy of polycythemia vera is 74–111 MBq/m$^2$ body surface area with a maximum activity of 185 MBq.

The ICRP assumes that 30% of administered $^{32}$P phosphate is permanently accumulated in the mineral bone and 30% is distributed in soft tissues, where it is excreted with half-times of 12 h (20%), 2 days (20%) and 19 days (60%). The absorbed dose for bone surfaces is 11 mGy MBq$^{-1}$ and other organs receive 0.74 mGy MBq$^{-1}$. The effective dose for $^{32}$P phosphate is approximately 2.2 mSv MBq$^{-1}$.

**Related Article:** Phosphorus-32


**PA (posteroanterior) projection**
(General) See Posteroanterior (PA) projection

**PACS (picture archiving and communication systems)**
(Nuclear Medicine) See Picture archiving and communication systems (PACS)

**Pair production**
(General) Pair production dominates photon interaction processes at high energies (above 50 MeV). The incident photon interacts with the nuclear field. The entire photon energy is absorbed to produce an electron–positron pair. As the rest mass of an electron or positron is equivalent to 511 keV, there is an energy threshold for pair production of 1.022 MeV.

The emitted positron quickly annihilates with an electron within the medium, producing two photons each of energy 511 keV. Above 2.044 MeV, triplet production may occur. In this case, the photon interacts with the field of an electron, giving rise to an electron–positron pair plus the interacting electron.

The cross section for pair production varies with $Z^2$. There is also a weak dependence on $E$.

**Related Article:** Electron–positron pair

**Palliative treatment**
(Radiotherapy) Where cure from radical treatment is unlikely, palliative treatment can alleviate painful or distressing symptoms and restore a higher degree of life quality for patients. Palliative treatments tend to deliver lower doses than radical treatments but can have a most favourable impact to bring relief in the cases where disease has compressed nerves (cord compression) or other structures (oesophagus).

**Related Articles:** Parallel opposed fields, Beam arrangement

**Paper and thin layer chromatography**
(Nuclear Medicine) Chromatographic methods are very suitable for the separation of compounds of similar chemical structure. The appropriate chromatographic method has to be chosen according to the physicochemical properties of the compounds.

In paper chromatography, the solution of the compounds is put in a small droplet at the end of a specially prepared paper. After the spot is dried, the paper (with the end including the spot) is put in a suitable solvent. Different compounds can then be separated due to their different migration velocities.

Thin layer chromatography is similar to paper chromatography although a glass plate is used. The plate is prepared with a thin layer of adsorbent or porous material.

In liquid chromatography a solution of the compounds is dropped on the upper end of a column of adsorbent material. The column is developed by adding a solvent to the upper end of the column.

In the aforesaid methods the radioactive distribution is determined and a radio chromatogram is produced. The locations of the activity peaks can then be used for identifying the compounds in the sample. Prior to measurement, the system has to be calibrated with known samples.

**Abbreviations:** PC = Paper chromatography, TLC = Thin layer chromatography, LC = Liquid chromatography and HPLC = High-performance liquid chromatography.


**Paraffin phantom**
(General) A tissue-equivalent phantom for medical imaging and therapy, made from paraffin wax.

**Uses in Medicine:** Dosimetry – Paraffin phantoms can be used in the estimation of absorbed dose in a region of the body resulting
from external exposure to ionising radiation. An appropriate radiation sensitive material (radiographic film, OSL or TLD material, etc.) is placed within the phantom and the phantom is exposed. The total dose to a point from the exposure can be read from the radiation sensitive material in the usual manner. Since the phantom has similar attenuation and scatter characteristics to tissue, this dose is assumed to be representative of that which would be given to a tissue or organ located in the equivalent position in a patient.

Radiotherapy – Paraffin phantoms can also be used in Radiotherapy to compensate for ‘missing tissue’. Radiation travelling through the phantom will exit with a modified energy spectrum and spatial distribution, allowing the use of a standard Radiotherapy treatment plan when oblique beams are used, or when the beam must pass through an inhomogeneous medium or an uneven skin surface.

Parallel acquisition technique (PAT)

(Magnetic Resonance) Parallel acquisition technique (PAT) is a vendor term (Siemens) for partial parallel imaging, i.e. MRI acquisition techniques using multiple receiver coils to allow k-space undersampling. The receive profiles of the coils must be spatially dependent and complement each other in at least one direction. If these are known, the origin of the signals can be reconstructed from the (otherwise aliased) MR images (SENSE) or the undersampled k-space signal (SMASH).

Related Articles: SMASH, SENSE

Parallel imaging

(Magnetic Resonance) Parallel imaging is a technique enabling faster MRI scanning without the need to reduce image resolution. In conventional MR imaging, the number of lines in k-space acquired must equal the number of voxels in the phase encoding direction. For a fixed field of view, each line added in the phase encode direction improves resolution but costs time. Reducing scan time by skipping the acquisition of lines in k-space (Figure P.1) results in undersampling of spatial frequencies. On transformation to the spatial domain, undersampling becomes apparent as fold over or ‘aliasing’ of the image.

Parallel imaging allows acquisition of a reduced number of k-space lines, while avoiding the aliasing problem. The acceleration in scan time achieved is given by the factor $R$. Parallel imaging employs multiple coil elements, each with a known spatial sensitivity to signal (Figure P.2).

There are two methods of reconstructing an image using parallel imaging. In SENSE imaging (Figure P.3), reconstruction takes place after the Fourier transform, in the spatial (image) domain. Data is collected simultaneously from all coil elements and undersampled k-space is filled for each coil. A Fourier transform generates an aliased image from each coil. As coil spatial sensitivities differ, each image generated will be unique. If the spatial sensitivity of a given coil $j$ is $C_j(x,y)$, then the image $I_j(x,y)$ for the $j$th coil element of $N$ coils is

$$I_j(x,y) = C_j(x,y)ρ(x,y) + C_j(x,y+Δy)ρ(x,y+Δy) + C_j(x,y+2Δy)ρ(x,y+2Δy) + \cdots
$$

where $ρ(x,y)$ represents signal intensity corresponding to point $(x,y)$ in the ideal, unaliased MRI image of the object. Each successive term aforesaid represents an alias term modulated by the coil sensitivity profile $C_j(x,y)$ for that particular coil $j$. If the number of alias terms equals the sense factor $R$. The function $ρ(x,y)$ is periodic over a spatial extent of the field of view (FoV) and the spatial shift in each term $Δy$ is $\text{FoV}/R$.

Each of the $N$ coil elements generates an expression in the form of (P.1). Expressing in matrix notation

$$\begin{bmatrix}
I_1(x,y) \\
I_2(x,y) \\
\cdots \\
I_N(x,y)
\end{bmatrix} =
\begin{bmatrix}
C_1(x,y) & C_1(x,y+Δy) & \cdots & C_1(x,y + RΔy) \\
C_2(x,y) & C_2(x,y+Δy) & \cdots & C_2(x,y + RΔy) \\
\cdots & \cdots & \cdots & \cdots \\
C_N(x,y) & C_N(x,y+Δy) & \cdots & C_N(x,y + RΔy)
\end{bmatrix}
\begin{bmatrix}
ρ(x,y) \\
ρ(x,y+Δy) \\
\cdots \\
ρ(x,y + RΔy)
\end{bmatrix}
$$

which can be simplified to:

$$I = C\bar{ρ}
$$

where

$I$ and $\bar{ρ}$ are vectors

$C$ is a matrix representing the coil sensitivities

The expression can be solved for the unaliased image data $\bar{ρ}$ once the number of coils $N$ is greater than or equal to the number of ‘unknowns’ in the rows of the vector $\bar{ρ}$. That is, the number of coil elements must equal or exceed the acceleration factor $R$. It is also necessary that the individual coil spatial sensitivities are sufficiently different over the volume of interest.
In an alternative realisation of parallel imaging called ‘SMASH’ (also called GRAPPA), the ‘missing’ $k$-space lines are filled prior to image reconstruction by the FFT. For an individual coil, the $k$-space value (i.e. the signal intensity) corresponding the spatial frequencies $k_x, k_y$ is

$$S(k_x, k_y) = \int \int \rho(x,y) \exp(-ik_x x, -ik_y y) dx dy$$  \hspace{1cm} (P.4)

By weighting individual coil elements, $j$ by a weight $w_j$, the overall profile $C(x,y)$ can be controlled. With weightings set to $w_j^0$ to give an overall uniform sensitivity profile (i.e. combined $C(x,y) = 1$) the lines in $k$-space are filled with values $S(k_x, k_y)$:

$$S(k_x, k_y) = \sum_{j=1}^{N} w_j^0 S(k_x, k_y)$$  \hspace{1cm} (P.5)

Substituting (P.4) into (P.5)

$$S(k_x, k_y) = \sum_{j=1}^{N} \int \int w_j^0 C_j(x,y) \rho(x,y) \exp(-ik_x x, -ik_y y) dx dy$$

$$= \int \int \rho(x,y) \exp(-ik_x x, -ik_y y) dx dy$$  \hspace{1cm} (P.6)

From the definition of the Fourier transform, this represents the FT of $\rho(x,y)$ as in (P.6) mentioned previously, but shifted in spatial frequency by $\Delta k_y$. The missing lines in $k$-space can then be constructed through weighted combination of the signals from each coil as

$$S(k_x, k_y + \Delta k_y) = \sum_{j=1}^{N} w_j^0 S(k_x, k_y)$$ \hspace{1cm} (P.9)

For both SENSE and SMASH techniques the coil sensitivities $C_j(x,y)$ must be known. The sensitivities can be measured in a prescan or in some variants of the techniques by acquiring additional calibration $k$-space line during scanning.

Implementation of parallel imaging reduces SNR by minimum factor of $\sqrt{R}$, all other factors remaining equal. In practice, the SNR reduction with parallel imaging is somewhat worse due to imperfections in coil geometries, given by a geometry factor '$g'$:

$$\text{SNR}_{\text{parallel}} = \frac{\text{SNR}}{g\sqrt{R}}$$  \hspace{1cm} (P.10)

With parallel imaging, SNR is variable across the image.

Parallel opposed fields
(Radiotherapy) This form of treatment involves two coaxial beams entering the patient at 180° to one another (Figure P.4). It is typically used where it has not been possible to accurately define the tumour, or for palliative treatments involving a low dose. These treatments tend to be relatively straightforward plans to calculate, and a manual calculation may be performed with the dose distribution being calculated from simple data tables and knowledge of the separation between the two beam entry points on the skin surface. The dose is generally prescribed to the midpoint of the target volume.

The summation of the two opposing percentage depth dose curves yields a plateau-shaped region as demonstrated in Figure P.5 for two 6MV beams separated by 20 cm.

Related Articles: Palliative treatment, Beam arrangement, Geometric field separation

Parallel organs
(Radiotherapy) Radiation treatment inevitably affects normal tissue and so may cause radiation-induced adverse effects. The tolerance of normal tissues to radiation depends on the ability of the clonogenic cells to maintain a sufficient number of mature cells suitably structured to conserve organ function. The tissue architecture is thought to be important in determining the tolerance dose for partial organ irradiation.

In radiotherapy, it is generally the case that the total dose that can be tolerated depends on the volume of tissue irradiated – the dose–volume effect. It has been suggested that groups of cells within an organ are organised into collective bodies called functional subunits (FSU). The arrangement of the FSUs within the tissue is thought to be an important factor in determining the volume dependence of an organ. In parallel organs, the FSUs are arranged in parallel so the inactivation of a small number of FSUs does not lead to loss of organ function. Inactivation of a critical number of FSUs is required for functional damage to occur, meaning that there should be a threshold volume of irradiation below which no functional damage will develop even after high-dose irradiation. Above this threshold there is a graded rather than a binary response: Functional impairment increases in severity with increasing dose.

When planning radiotherapy treatment, assessment is always made of the dose to normal tissues and modern commercial treatment planning systems have a number of tools to aid this process including dose–volume histograms (DVH). For parallel organs, complications are likely to be dependent on the average dose to the whole volume. Reduction of the volume of normal tissue irradiation increases the dose that can be received without inducing an adverse effect. For some organs this is quite a significant effect. For example, Emami suggests that for the kidney reducing the volume irradiated from the whole organ to only a third increases the tolerance dose from 23 to 50 Gy.

Examples of parallel organs are kidney, lung, liver. It should be noted however that most real normal tissues have a mixed parallel and serial architecture.

Abbreviations: FSU = Functional sub-unit and DVH = Dose–volume histogram.

Related Articles: Adverse events, Dose volume histogram, Dose response model, Serial organs, Sigmoid dose–response curve, Tolerance


Parallel plate ionisation chamber
(Radiotherapy) Ionisation chambers are usually with cylindrical or parallel plates. The latter used in radiotherapy (akd plane parallel ionisation chambers) are designed mainly for low energy x-rays and electron beams that feature steep depth dose gradients in a homogeneous phantom. According to most dosimetry protocols, parallel plate chambers are recommended to be used at all electron energies, and below 10 MeV their use is mandatory. For photon beams they are suitable for reference dosimetry measurements only when a calibration in terms of absorbed dose to water is available at the user quality. They are also suitable for reference dosimetry for proton and heavy ion beams. In high energy photon beams plane-parallel chambers are useful for measurements in the dose build-up region. Figure P.6 shows a diagram of a parallel plate ionisation chamber.

The design of a chamber with parallel plates is usually more complex than that of cylindrical chamber due to the presence of a guard electrode and a large amount of back-scatter material used in the chamber construction. A guard electrode is an important element of the construction of a parallel plane chamber because it is useful not only to obtain a homogeneous electrical field in the sensitive volume of the chamber (Figure P.7) but also to avoid the secondary electrons scattered from the chamber wall being counted in the chamber.

The electrical field lines of plane parallel ionisation chamber are parallel to the direction of the incident radiation (Figure P.8). The secondary electrons are predominantly forward directed and therefore they can gain energy if the entrance window is negatively charged and the collecting electrode is positively charged. This may lead to more ionisation events than in the opposite polarity. Therefore in plane parallel ionisation chambers the polarity effect is more pronounced than in cylindrical chambers. Parallel plane chambers are generally not well suited for scanning measurement in water because they create disturbances in the water during their movements due to their shape and it is difficult to set water surface level accurately due to the adhesion to the water meniscus. Therefore the parallel plane chambers are more commonly used in combination with solid slab phantom. The characteristics of some parallel plane chambers are given in Table P.1.

Related Article: Ionisation chamber
Parallel-hole collimators


Parallel-hole collimators
*(Nuclear Medicine)* A specific collimator design used to attain a spatial resolution with a scintillation camera. The collimator consists of a high absorbing material with parallel holes drilled or cast in it. Photons with propagation near-parallel to the holes direction will pass through and reach the detector. Photons with an oblique angle of incidence will most likely be absorbed when passing through the lead septa separating two holes and will therefore be prevented from reaching the detector. A parallel-hole collimator will produce an image with the same size and orientation as the object size.

By altering the length and width of the holes it is possible to construct collimators with different features, e.g. long holes with a small width will produce an image with high spatial resolution but low count rate since a lot of photons will be absorbed by the collimator and vice versa when having short broad holes.

It is necessary to increase the septa width when using radionuclides emitting high energy photons to prevent septal penetration. Thick septa on the other hand will absorb too much of the incident radiation and therefore decrease the collimator efficiency. Septal penetration yields a decrease in spatial resolution and it is therefore important to consider the photon energy when selecting a collimator. The parameters for a number of different collimator types can be seen in Table P.2.

A collimator is one of the degradable factors affecting the system spatial resolution. The contribution from a parallel-hole collimator is

\[
R_{\text{coll}} \approx d \left( \frac{l_{\text{eff}} + b}{l_{\text{eff}}} \right)
\]  

(P.11)

where
- \( b \) is the source to collimator distance
- \( d \) is the hole diameter
- \( l_{\text{eff}} \) is the effective length of the holes, which includes septal penetration

The spatial resolution is best when the source is located as close to the collimator face as possible and when the holes are long and thin (small width).

The geometric efficiency is defined as the fraction of incident photons registered by the detector. When using a parallel-hole collimator the geometric efficiency, \( g \), is expressed as follows:

\[
g \approx K^2 \left( \frac{d}{l_{\text{eff}}} \right)^2 \left( \frac{d^2}{(d + t)^2} \right)
\]  

(P.12)

where
- \( K \) is a constant depending on the hole shape
- \( t \) is the septal thickness

The geometric efficiency is not dependent on the distance between the source and the collimator face because even if the individual hole efficiency is decreased by \( 1/b^2 \) the irradiated detector area is increased by \( b^2 \) so the total count rate is unchanged. But since the spatial resolution is dependent on the source to collimator distance it is important to place the camera as close to the object as possible.

Related Articles: SPECT, Collimator, Parallel-hole collimator, Diverging collimator, Converging collimator, Collimator, Collimator design, Collimator parameters
TABLE P.1
Some Characteristics of Ionisation Chambers

<table>
<thead>
<tr>
<th>Ionisation Chamber Type</th>
<th>Materials</th>
<th>Window Thickness</th>
<th>Electrode Spacing (mm)</th>
<th>Collecting Electrode Diameter (mm)</th>
<th>Guard Ring Width (mm)</th>
<th>Recommended Phantom Material</th>
</tr>
</thead>
<tbody>
<tr>
<td>NACP01 (scanditronix)</td>
<td>Graphite window, graphited rexolite electrode, graphite body (black wall), rexolite housing</td>
<td>4 mm–90 mg/cm²</td>
<td>2</td>
<td>10</td>
<td>3</td>
<td>Polystyrene, graphite, water (with waterproof housing)</td>
</tr>
<tr>
<td>NACP 02 (scanditronix)</td>
<td>Mylar foil and graphite window, graphite rexolite electrode, graphite body (black wall), rexolite housing</td>
<td>0.6 mm–104 mg/cm²</td>
<td>2</td>
<td>10</td>
<td>3</td>
<td>Water, PMMA</td>
</tr>
<tr>
<td>Markus chamber PTW</td>
<td>Graphited polyethylene foil window, graphited polystyrene collector, PMMA body, PMMA cap</td>
<td>0.9 mm–102 mg/cm² (incl. cap)</td>
<td>2</td>
<td>5.3</td>
<td>0.2</td>
<td>Water, PMMA</td>
</tr>
<tr>
<td>Scedx–Welhofer PPC 05</td>
<td>Window and body C-552, graphited (PEEK) electrode</td>
<td>1 mm–176 mg/cm²</td>
<td>0.5</td>
<td>10</td>
<td>3.5</td>
<td>Water</td>
</tr>
<tr>
<td>Holt chamber (memorial)</td>
<td>Graphited polystyrene wall and electrode, polystyrene body</td>
<td>4 mm–416 mg/cm²</td>
<td>2</td>
<td>25</td>
<td>5</td>
<td>Polystyrene (phantom integration)</td>
</tr>
<tr>
<td>Capintec PS-033</td>
<td>Aluminised mylar foil window, carbon impregnated air equivalent, plastic electrode, polystyrene body</td>
<td>0.004 mm–0.5 mg/cm²</td>
<td>2.4</td>
<td>16.2</td>
<td>2.5</td>
<td>Polystyrene (phantom integration)</td>
</tr>
<tr>
<td>Exradin 11</td>
<td>Conducting plastic wall and electrodes Model P11: polystyrene equivalent Model A11: C-2552, air equivalent Model T11: A-150 tissue equivalent</td>
<td>P11: 1 mm–104 mg/cm²</td>
<td>2</td>
<td>20</td>
<td>5.1</td>
<td>P11: polystyrene water</td>
</tr>
<tr>
<td>Roos chamber PTB FK6</td>
<td>PMMA, graphited electrodes</td>
<td>1 mm–118 mg/cm²</td>
<td>2</td>
<td>16</td>
<td>4</td>
<td>Water PMMA</td>
</tr>
<tr>
<td>PTW 34001 Scedx</td>
<td>Kapton conductive film window, graphited polyethylene collector, solid water body</td>
<td>0.025 mm–4.8 mg/cm²</td>
<td>1 mm (0.7 mm reported)</td>
<td>12.7</td>
<td>13.5</td>
<td>Solid water</td>
</tr>
<tr>
<td>Attix chamber RMI 449</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Source: Data courtesy to the manufacturers listed in the first column.


**Paralysable counting system**  
*(Radiation Protection)* The dead time $\tau$ in a detector is a minimum time interval which must separate two events to be measured (registered) as two separate pulses.

In a paralysable counting system the events that occur during the dead time period are not just missed but will also restart the dead time, so that with increasing event rate the detector will reach a saturation point where it will be incapable of recording any event at all. The measured rate $M$ can be expressed as equal to

$$M = N \times \exp(-N \times \tau)$$

where

$N$ is a true interaction rate

$\tau$ is a dead time of the detector system

Related Articles: Deadtime losses, Non-paralysable counting system


**Paramagnetic contrast agents**  
*(Magnetic Resonance)* Agents based on paramagnetic ions are the most common form of contrast agent used in MRI. A number of different ions have been proposed (e.g. Mn$^{2+}$, Dy$^{3+}$). However, gadolinium (Gd$^{3+}$) remains the mainstay of commercially available agents.
Related Articles: Contrast agent, Gadolinium chelate, Positive contrast media

Paramagnetism (Magnetic Resonance) In a paramagnetic material, the electrons are unpaired and bulk magnetic properties are only apparent in the presence of an externally applied field. When the paramagnetic material is removed from the externally applied field no permanent magnetisation exists in the material. Paramagnetic materials have a positive magnetic susceptibility $\chi > 0$. Magnetic field is strengthened by a paramagnetic material. Examples of paramagnetic materials are: aluminium [13], barium [56], calcium [20], gadolinium [31] and magnesium [12]. The numbers in the parenthesis are the atomic number of the material.

Gadolinium in a chelate form is often used as a contrast agent in MR imaging because of its $T_1$-shortening effect. Gadolinium by itself is toxic. Examples of gadolinium chelates are gadopentetate dimeglumine (Gd-DTPA) and gadoterate meglumine (Gd-DOTA).

Related Articles: Susceptibility, Ferromagnetism

Parent radionucleus (Nuclear Medicine) The first radioisotope in a subjectively chosen decay chain. In a decay chain the initial radioisotope disintegrates into another radioactive isotope. The second radioisotope is called the daughter radioisotope, progeny or decay product. The daughter radioisotope will eventually decay to a second decay product, or grand-daughter radioisotope. This chain will continue until one of the decay products is stable. In nuclear physics, a specific part of a decay chain can be of clinical and/or research interest (the parent radioisotope can be chosen subjectively). For instance $^{99}$Mo disintegrates to the meta-stable state of $^{99m}$Tc. The parent radionuclide in this example is $^{99}$Mo. The relations between parent and daughter activation is described by the Bateman equations.

Parent–daughter decay
(Nuclear Medicine) Parent–daughter decay refers to the situation in which a radioactive daughter nucleus is produced by a decaying parent nucleus. A parent–daughter pair commonly used in nuclear medicine is $^{99}$Mo and $^{99m}$Tc where $^{99}$Mo decays to $^{99m}$Tc. $^{99m}$Tc is continuously produced and eluted from a generator containing $^{99}$Mo and used for imaging applications. If the $^{99m}$Tc is not eluted regularly the activities of the two radionuclides will reach a transient equilibrium.

A parent–daughter can be modelled by the Bateman equations which describe a decay sequence with two or more radionuclides:

$$A_d = \left[ A_p(0) \frac{\lambda_d}{\lambda_d - \lambda_p} \times \left( e^{-\lambda_d t} - e^{-\lambda_p t} \right) \right] \times BR + A_d(0) e^{-\lambda_d t}$$

where

- $A_p(t), A_d(t)$ are the activities of the parent and the daughter respectively at time $t$
- $\lambda_c, \lambda_d$ are their respective decay constants
- $BR$ is the branching ratio for situations where there is more than one decay modes for the parent

The last term in Equation P.13 is the residual daughter-product activity that might be present at time $t$.

For a $^{99}$Mo-$^{99m}$Tc generator $t = 0$ is set to the first elution; hence, the daughter activity at $t = 0$, $A_d(0)$ can be assumed to be zero.

Related Article: Transient equilibrium

Pareto chart (Nuclear Medicine) A special kind of bar chart where the plotted values are arranged in descending order. The bars are usually accompanied by a line graph that represents the cumulative value of the bars. The Pareto chart is used as one of the many tools in the quality assurance process.

Paris system (Radiotherapy, Brachytherapy) The Paris system is a predictive dosimetry system for interstitial brachytherapy originally developed for $^{192}$Ir wires. This system involves the use of rules and has proved to be simple, reliable and clinically efficacious.

The Paris system is based on the following principles:

- Sources must be straight, parallel and arranged so that their midpoints are located in the same plane, perpendicular to the line sources. This plane is called the central plane, and it is the mid-plane of the application.
- The reference linear air kerma rate, the apparent linear activity, along each line source must be uniform and identical for all wires.
- Adjacent line sources must be equidistant. If the volume to be treated is large, then it is necessary to perform implants in more than one plane, the separation between planes being such that the principle of equidistant line sources is observed.
Brachytherapy generally produces a very heterogeneous dose distribution, with high doses close to the sources, and lower doses between them. A careful arrangement of the sources, following the rules in the Paris system, results in a relatively uniform dose between sources, and these dose values are used as reference values for the evaluation of the dose delivered to the tumour. Thus, the dose specification is based on dose values inside the target volume in the central plane.

The basal dose rate, BD, is defined in the central plane as the mean value of the minimum dose rates, the elementary basal dose rates, BD_i:

$$BD = \frac{BD_1 + BD_2 + \ldots + BD_n}{n}$$

For a single plane implant with four wires (Figure P.9), the BD_i is calculated at the midpoint of segments joining the intersections of the line sources with the central plane.

For volume implants with five wires (Figure P.10), the elementary basal dose rates are calculated ‘in the triangle midpoints’; defined by perpendicular bisector lines projected from the sides of the triangles formed by the intersections of the line sources with the central plane. For a perfect square, the midpoint is used.

In clinical situations these ideal source patterns are difficult to realise. The implant is acceptable if the elementary basal dose rates satisfy the following relationship (and no triangles have any obtuse angles):

$$0.9 \, BD \leq BD_i \leq 1.1 \, BD$$

Dose is specified in relation to the basal dose, and the reference isodose rate (RD) surface used in the Paris system corresponds to 85% of the basal dose:

$$RD = 85\% \, BD$$

The reference isodose should encompass the target volume as closely as possible. The treated volume is the volume encompassed by the reference isodose surface. The high dose volume, ‘the hyperdose sleeve’, is the region around each wire receiving more than twice the reference dose. It is to be noted, that the Ir-wires extend outside the treated volume for a Paris system implant, and the ratio of treated length to active wire length is 0.7.

The Paris system has been used for many years with good clinical results. Originally developed for 192Ir-wires, the system can be used also for 192Ir seed ribbons and for afterloading units with a stepping source when a standard step size and equal dwell times are used. Modifications based on the Paris system have been introduced for stepping source systems, taking advantage of the possibility to optimise the dwell times.

**Abbreviations:**
- ADC = Analogue digital converter
- EPI = Echo planar imaging
- OPG = Ortho pan tomography
- PPI = Partial parallel imaging
- PFI = Partial Fourier imaging

**Partial Fourier imaging (PFI)**
(Magnetic Resonance) Partial Fourier imaging (PFI) is a partial acquisition technique in MRI that exploits the fundamental Hermitian symmetry in k-space when magnitude images are of interest. Contrary to partial parallel imaging, it does not require the use of phased array receive coils.

A certain fraction of phase encoding lines is not acquired (starting from high-spatial frequencies, but clearly above 50%). The missing data points are extrapolated from the central k-space signals using specific algorithms (for example Margosian, POCS). Under-representation of low-SNR spatial frequencies will result in deterioration of the point spread function and increased SNR, especially with zero-filling.

**Partial parallel imaging (PPI)**
(Magnetic Resonance) Partial parallel imaging (PPI) is the general term for MRI acquisition techniques using multiple receiver coils to allow k-space undersampling. The receive profiles of the coils must be spatially dependent and complement each other in at least one direction (as expressed by the g-factor). If these are known the origin of the signals can be reconstructed from the (otherwise aliased) MR images (SENSE) or the undersampled k-space signal (SMASH).

PPI cannot be performed in a direction along which the coils’ sensitivity changes in the same manner, e.g. axial for head phased array and transverse for spine phased array.

PPI is performed in one of the phase-encoding directions to allow for shorter measurement times (spin-warp) or echo-trains (EPI). It comes at a penalty of reduced signal-to-noise ratio (SNR). Depending on the design of the coil array and direction, the reconstruction problem becomes ill-conditioned at some degree of undersampling, resulting in image artefacts.

**Partial volume effect**
(Diagnostic Radiology) In CT, the partial volume effect (artifact) occurs when a high contrast structure extends only partly into the imaged slice. The CT numbers of the pixels should represent the average attenuation of the voxel within the slice. However, due to the logarithmic relationship between intensity and attenuation values, the calculated average CT number of the pixels will not be a true representation of the materials within the voxels (Figure P.11).
The partial volume effect results in a reduced contrast (Figure P.12a). It can be eliminated, and contrast optimised, by using a narrower imaged slice thickness (Figure P.12b).

The partial volume effect can also result in artefacts which appear as streaking in the image. This will be the case for off-centre objects projecting slightly into the scan plane. Due to beam divergence, inconsistencies will occur between views obtained from opposing directions, which give rise to streaking artefacts. When the x-ray tube is at 0° the high density object is in the path of the beam but with the tube at 180°, the object is not in the path of the beam (Figure P.13). The partial volume artefact will also be reduced by use of a narrower imaged slice thickness.

Related Articles: Streaking artefact, Slice thickness

Particle radiation

(Radiation Protection) The term ‘particle radiation’ is normally used to differentiate from electromagnetic radiation, although photons can be considered as particles. The particles involved can be charged or uncharged. The term encompasses not only particles such as protons, neutrons and electrons but also ions.

Related Articles: Beta radiation, Beta+ radiation, Photons, Radiation, Alpha particle

Particle therapy

(Radiotherapy) Particle therapy is an alternative name for Hadron therapy. See article Hadron therapy for more information.

Related Article: Hadron therapy

Particle velocity

(Ultrasound) When an ultrasound wave propagates in a medium, the particles within the medium will start oscillating. The displacement is the particle’s distance, at a certain time, from its rest position. A transducer surface moving in a sine-shaped movement will give rise to sinusoidal motions of the particles with the velocity \( v \). The relations between displacement, particle velocity and pressure for such a wave (expressed in 1D) are

\[
\begin{align*}
\text{Displacement: } s &= s_0 \cos(\omega t - kx) \\
\text{Particle velocity: } v &= \partial s/\partial t = -\omega s_0 \sin(\omega t - kx) \\
\text{Pressure: } P &= Zv = -Z\omega s_0 \sin(\omega t - kx)
\end{align*}
\]

If the pressure amplitude \( p \) and the acoustic impedance are known, the particle velocity amplitude (\( \omega s_0 \)) can be calculated as

\[
\omega s_0 = p/Z = 10^6 \text{Pa}/1.6 \times 10^6 \text{Rayls} = 0.6 \text{m/s}
\]

Related Articles: Displacement, Pressure
Partition coefficient

(Nuclear Medicine) In tracer kinetics a compartment may be open or closed to a tracer. For example when tracers are able to move between the two compartments of blood and tissue, these compartments are open. After a given period of time the activity ratio in the blood and tissue reaches a steady state. The ratio of tissue activity concentration $C_t$ (Bq/g) to blood concentration $C_b$ (Bq/mL) at this point is known as the partition coefficient $\lambda$ (mL/g).

The partition coefficient $\lambda$ (in a steady state system) is defined by

$$\lambda = \frac{C_t}{C_b}\quad(P.14)$$

$C_b$ can be directly measured by using blood samples.

If one assumes that the tracer concentration in the blood is equal to the concentration in tissue, the apparent distribution volume is given by

$$V_t = \frac{A_t}{C_t}\quad(P.15)$$

$$C_b = \frac{A_t}{V_t}$$

where $A_t$ is the activity in the tissue. If the concentration in the blood is lower than the concentration in the tissue the apparent distribution volume $V_t$ will be smaller than the actual distribution volume $V_t$ and vice versa if the tracer concentration is higher. The distribution volume of the tissue equals the ratio between the activity and the concentration which is given by

$$C_t = \frac{A_t}{V_t}\quad(P.16)$$

Combining these two relationships (Equations P.15 and P.16) and Equation P.14 yields

$$\lambda = \frac{V_t}{V_t}\quad(P.17)$$

Therefore the partition coefficient can also be interpreted as the distribution volume per unit mass of tissue for a tracer, which is convenient when estimating blood flow and perfusion.

Related Articles: Tracers, Analogue tracers, Distribution volume


Passive device

(Magnetic Resonance) The term passive implant refers to any medical device that serves its function without the supply of power. Examples of passive implants include but are not limited to dental implants, ocular implants, orthopaedic implants and penile implants. Some metallic implants have shown considerable torque when placed in the presence of the strong static magnetic field. The force or torque exerted on small and large metallic implants can cause serious effects as unanchored implants can potentially move unpredictably within the body. The mass and the type of metal used in such implants is one factor which determines the force exerted on them in the static magnetic fields. While non-ferrous metallic implants may show little or no deflection to the field they could cause patient burning because of the significant heating due to their inability to dissipate the heat caused by RF absorption.

Passive shielding

(Magnetic Resonance) Access to the static magnetic fringe field is restricted within the so-called pacemaker safety limit (for magnetic fields above 0.5 mT), extends 5–10 m (depending on design) from a completely unshielded high field strength MRI magnet. Therefore most clinical installations require a degree of shielding. Usually the magnet is equipped with active shielding (see Related Article). However, passive shielding is possible and is obtained by use of ferromagnetic materials, most commonly iron.

The iron can be integral to the magnet housing (self shielding) but is usually mounted in the walls/roof/floor of the scanner room. If passive shielding is used in combination with active shielding, partial shielding of the room is often sufficient. This may be the case when the 0.5 mT limit otherwise would extend outside the scanner room (requiring access to be restricted) or if there is equipment in an adjoining room that is very sensitive to static magnetic fields, such as a gamma camera or a PET scanner.

The iron shielding works due to its magnetic permeability:

$$\mu = B/H\text{ [H/m]}$$

where

- $B$ is the magnetic flux density [T]
- $H$ is the magnetic field strength [A/m]

This can be described as the material’s ability to ‘concentrate’ magnetic fields. The passive shielding thus provides a more preferable return path for the magnetic flux around the magnet than the air would and limits the extent of the static magnetic field. The so-called magnetisation curve is shown schematically in Figure P.14, showing $B$ as a function of $H$. Eventually, for a sufficiently high $B$, the material saturates and cannot further ‘concentrate’ (or shield) the static magnetic field.

Other materials with very high permeability, such as permalloy, could be an alternative but they are expensive and tend to saturate at lower magnetic flux densities than iron.

Related Articles: Active shielding, Fringe field

Further Reading: Krestel, E. 1990. Imaging Systems for Medical Diagnostics, Siemens Aktiengesellschaft, Berlin, Germany.

PAT (parallel acquisition technique)

(Magnetic Resonance) See Parallel acquisition technique (PAT)
Peak areas

Fundamentally, the area under a peak in an NMR spectrum reflects the quantity of spins in the corresponding chemical group within the region from which the signal has been acquired. However, the peak area peak may also be dependent upon a number of other factors: The NMR visibility of nuclei may be affected by physicochemical properties such as binding. Peaks due to a specific chemical group may be split due to coupling. Importantly, in a conventional NMR experiment, peak area may be weighted according to the equipment couch or stand. For example, erect lateral chest x-ray.

Naturally this approach has now been replaced by computer-based methods, but many of the same problems remain, such as how to allocate areas between overlapping peaks, and how to handle broad baseline features. Sophisticated model-based fitting packages, operating in either the time domain or the frequency domain, allow the user to tackle these issues and to assess the quality of the resulting fit (Figure P15).

Peak assignment

Peak assignment refers to the process by which the resonance peaks in an NMR spectrum are attributed to specific chemical groups and compounds. For in vivo MRS, it relies heavily on prior knowledge.

Historically, peak assignment in in vivo spectroscopy has been a difficult task. It is far from trivial in NMR spectroscopy in general, since there is no one-to-one relationship between resonance peaks and specific chemical compounds. A peak may contain contributions from a number of compounds containing the nucleus of interest in the same chemical environment, while compounds that contain the nucleus in more than one chemical group give rise to multiple peaks in the spectrum.

A high field spectrum may contain resolvable signals from dozens, if not hundreds, of different compounds. At the field strengths used for in vivo work, there is a further complication in that these peaks collapse into a small number of composite, often overlapping features. Establishing the composition of these features can be a formidable task; particularly as the proportions of different compounds contributing to a peak may vary with tissue type, age and pathology. Peak assignment in 31P spectroscopy is arguably somewhat more straightforward than in 1H spectroscopy, although complicated by such phenomena as pH-dependent chemical shifts and the broad phospholipid baseline.

Peak assignment has relied heavily on a combination of prior knowledge of tissue biochemistry (i.e. what compounds are expected to be present in a specific tissue) and analysis of tissue extracts. For example, perchloric acid extracts can be used for analysis of low molecular weight, water soluble metabolites and chloroform-methanol extraction for lipids. Comparison of high-resolution
NMR spectra of these different extracts can provide clues as to the identity of specific components. Comparison of the more detailed spectra with the results of in vivo studies gives insight into the make-up of composite peaks, and finally assignment can be confirmed by comparison with spectra from pure metabolites in solution. Alternatively the extracts can be subjected to other forms of chemical analysis to identify the components.

**Peak kilovoltage (kVp)**

*(Diagnostic Radiology)* The peak kilovoltage is the maximum, or peak, voltage that is applied to an x-ray tube during the duration of the exposure, or during the electrical power supply cycle as illustrated in Figure P.16.

The electrical potential (voltage) applied to an x-ray tube might change during the exposure interval, with the waveform depending on the type of high-voltage power supply or generator being used. The technique factor value that is set and displayed for a specific clinical procedure is the peak value, the kVp. This is a quantity that can be measured when calibrating x-ray equipment.

The significance of the instantaneous voltage is that it determines the efficiency of x-ray production that varies throughout the cycle.

The significance of the effective voltage is that it determines the rate of heat production.

With constant potential generators, where the voltage does not vary during the exposure, the peak, effective and instantaneous values are the same. With sinusoidal power supply the peak value is equal to the effective value multiplied by \( \sqrt{2} \). For example, with an effective voltage of 220 V the peak value is 311 V.

**Peak scatter factor (PSF)**

*(Radiotherapy)* The peak scatter factor is the ratio of \( D_p \), the absorbed dose in tissue at the depth of dose maximum, to \( D_p^* \), the absorbed dose due to primary radiation only. The absorbed dose due to primary radiation is measured with just enough material to provide electronic equilibrium (i.e. ionisation chamber with build-up cap). It is defined as follows:

\[
PSF(A,E) = \frac{D_p(d_{max},A,d,E)}{D_p^*(A,E)}
\]

The measurement set-up is shown in Figure P.17. At low energies, \( d_{max} = 0 \) and PSF is known as the back-scatter factor.

There are two components to the behaviour of PSF with photon energy – the amount of back-scatter and the energy of the scattered photons. At low energy, there is a large amount of back-scatter but the energy is low, causing rapid absorption in the medium. As the energy increases the amount of scattering decreases, yet the photons have a higher energy and larger penetrating power. Hence the PSF increases with energy, reaching a maximum at a particular beam quality, and then decreases with further increases in beam energy. The PSF increases with field size, and may approach a saturation value (see Podgorsak 2003).

PSF is a special case of TAR (i.e. \( PSF(A_e) = TAR(d_{max},A_e) \)).

**Related Articles:** Percentage depth dose (PDD), Scatter air ratio (SAR), Tissue air ratio (TAR)


**Peak systolic velocity**

*(Ultrasound)* Systole describes the contraction of the chambers of the heart. The peak systolic velocity is the maximum velocity in the systemic arterial circulation produced by the contraction of the left ventricle (Figure P.18).

**Abbreviations:** PSV = Peak systolic velocity and EDV = End diastolic velocity.

**Related Article:** End diastolic velocity

**Peak voltage**

*(General)* The maximum value of an AC voltage, either positive or negative, measured from the point of reference. Peak voltage is only a moderately useful way of measuring AC voltage. The
most common way to measure/quantify AC voltage is the effective voltage or RMS voltage (see the eponymous article).

Related Articles: Effective energy, RMS voltage, Effective voltage value

Pencil beam

(Radiotherapy) A number of algorithms for computing the dose distribution similar to the ones used for photon beams have been applied for electron beams. Accurate planning with electron beams is complicated when the beams are incident on anatomical regions containing heterogeneous tissue. A common algorithm is based on the decomposition of a broad clinical beam into pencil beams each affected by the presence of small tissue inhomogeneities in their direct path. The conventional pencil beam algorithm was introduced by Hogstrom (1981). The concept of the Hogstrom pencil beam model is shown in Figure P.19. Each pencil beam has a size \((\Delta x, \Delta y)\), a mean direction \((\theta_x, \theta_y)\), an RMS angular spread about the mean direction \((\sigma_{\theta_x}, \sigma_{\theta_y})\), an energy \((E_{p,0})\) and a planar fluence \((\text{electrons cm}^{-2})\). A pencil beam is defined as all electrons passing through a pixel that is typically 2 mm\(^2\) at 100 SSD. Only the anatomy along the central axis of the pencil beam is used to calculate the dose distribution to the patient from that pencil beam. Four physical properties of the algorithm limit its accuracy: the slab geometry assumption, the planar fluence-base model, the Eyges Gaussian scatter distribution function and the inability to model the secondary electrons.


Penetrating radiation

(Radiation Protection) This is radiation which has sufficient energy to penetrate a material which it is incident on. X-rays, gamma rays and beta particles are types of penetrating ionising radiation. An obvious example of penetrating non-ionising radiation is light when incident on materials such as glass or water. The properties of the radiation (especially the wavelength/energy) and of the material itself determine how penetrating the radiation will be.

Penumbra

(Radiotherapy) The penumbra of a beam is normally defined as the distance between the 20% and 80% points of the dose profile, measured at 10-cm depth in a water phantom (Figure P.20). There are three components to the penumbra: transmission through the collimators, scatter penumbra and the geometric penumbra from the physical size of the source (i.e. physical penumbra). The geometric penumbra is the most significant contribution overall and is illustrated in Figure P.21.

For cobalt units, penumbras are generally 15-mm wide (cf. ~ 3–5 mm for LINACs). Reducing the distance \(b\) (as shown in Figure P.20) will reduce the penumbra width, but at the expense of increased scatter dose from the head components. The smaller the penumbra the more localised the delivered dose, enabling more accurate matching of fields.

Percentage depth dose

(Radiotherapy) A percentage depth dose curve, PDD, is commonly used in radiotherapy to graphically describe the dosimetric characteristics of a radiation beam. It shows the dose deposition

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**FIGURE P.20** Schematic of the beam profile showing the three regions.

**FIGURE P.21** Geometric penumbra, \(s\) – size of source.
Perception of depth, \( d \), in a particular medium, normalised to maximum dose at \( d_{\text{max}} \). The percentage depth dose calculation, for depth \( d \), field size \( A_s \) and source–surface distance (SSD), is given in Equation P.18. The field size parameter \( A_s \) refers to the value at SSD or at \( d_{\text{max}} \), not necessarily at the point of measurement:

\[
PDD(d, A_s, SSD) = 100 \frac{D_d}{D_{d_{\text{max}}}} = 100 \frac{D(d, A_s, SSD)}{D(d_{\text{max}}, A_s, SSD)} \tag{P.18}
\]

**Photon PDD:** A typical PDD for a MV photon beam in a tissue-equivalent material is shown in Figure P.22. The beam enters giving a surface dose \( D_s \), and as it penetrates the dose rises rapidly, reaching a maximum value at \( z_{\text{max}} \), after which it decreases almost exponentially. If the beam then exited the phantom, its value \( D_{\text{ss}} \) would be slightly lower than the extrapolated line due to the absence of scatter from points beyond the exit dose point.

The low surface dose (10%–30%) is known as the skin sparing effect, which is absent in orthovoltage or superficial beams. The subsequent build-up of dose results from the long range of secondary electrons released from the photon interactions, which carry energy deeper into the phantom. Charged particle equilibrium (CPE) is said to exist at \( z = z_{\text{max}} \), where \( z \) is approximately equal to the range of secondary charged particles, and is where the dose reaches its maximum. The dose then falls off due to photon attenuation, resulting in a transient CPE.

**Electron PDD:** A typical PDD for a MeV electron beam is a tissue-equivalent material shown in Figure P.23. The build-up is less pronounced for electron beams than for photon beams, due to the larger percentage surface dose for electrons (~80%). A single electron deposits the majority of its energy at the end of its track (a Bragg peak); however, the energy deposited along the central axis of a beam of electrons is more diffuse, a ‘smeared out’ Bragg peak. This is due to the oblique scattering of the electrons into the central axis. The surface dose increases with increasing energy (opposite to that of photons), due to the greater and larger angle electron scatter at low energies. At high energies, scatter is predominately forward. The dose then falls off due to electron attenuation, resulting in a transient CPE.

due to the electrons’ continuous energy loss and scatter. An additional consideration is the bremsstrahlung tail produced from the electrons interacting in the head, air and patient. The magnitude of this contribution depends on electron energy; typical contributions are less than 1% for 4MeV beams, and less than 4% for a 20-MeV beam. The strong scattering of electrons in air necessitates the use of applicators placed near the skin surface of patients, to help collimate the beam and reduce unnecessary dose outside the field.

**Perfusion imaging**

**Magnetic Resonance** In physiology, the term perfusion refers to the flow of blood through the capillary system. Perfusion or tissue blood flow (TBF) is traditionally quantified in terms of the volume of blood transported to a given mass of tissue per unit time, and a commonly used unit is mL/(min 100g). Other haemodynamic parameters such as blood volume and mean transit time (MTT) are sometimes loosely included in the term ‘perfusion parameters’. Additionally, a number of non- or semi-quantitative perfusion indices have been proposed in various applications. Perfusion imaging refers to the use of a medical-imaging modality for measurements of tissue blood flow. In quantitative approaches, the regional perfusion is normally visualised as a parametric map calculated on the basis of an appropriate tracer-kinetic model. Perfusion imaging is of particular interest for the assessment of cerebral blood flow (CBF) and myocardial blood flow (MBF), although investigations of numerous other organs or tissue types exist. A number of medical imaging modalities and techniques are used for the assessment of tissue perfusion, for example, SPECT, PET, MRI and CT.

**Myocardial Blood Flow:** Qualitative assessment of myocardial perfusion has traditionally been carried out by myocardial scintigraphy, that is by SPECT imaging of radiolabelled tracers such as TI-201
Perfusion imaging

(Thallium chloride) and Tc-99m-labelled sestamibi and tetrofosmin. Using these tracers, absolute quantification is prohibited by the fact that the myocardial uptake is not proportional to the myocardial blood flow (MBF) at high flow rates. Absolute quantification of MBF is feasible by use of PET in combination with H$_2^{15}$O as a freely diffusible tracer. MRI has emerged into an interesting alternative for myocardial perfusion imaging with excellent spatial resolution, primarily exploiting the first-pass tracer kinetics of conventional $T_1$-shortening extracellular gadolinium-based contrast agents. During the first passage, the contrast agent enters the microvasculature and starts to diffuse into the interstitial space, leading to increased signal intensity. Various imaging strategies exist, but a magnetisation preparation by saturation recovery followed by a steady-state free precession pulse sequence for readout is a common approach. Absolute quantification of MBF by cardiac MRI is indeed a challenging task, and it is fair to conclude that several problems remain.

**Cerebral Blood Flow:** Relative distributions of cerebral blood flow (CBF) are most commonly obtained by SPECT imaging of the intracellular retention of a Tc-99m-labelled tracer, such as hexamethyl propylenamino oxime (HMPAO) and ethyl cysteinolate dimer (ECD). A historically important diffusible tracer for regional CBF quantification in absolute terms is Xe-133. The Xe-133 tracer can be either inhaled or intravenously injected, the tracer kinetics is typically monitored by SPECT and CBF is calculated from modifications of the Kety–Schmidt tracer model. Somewhat similarly, the xenon-enhanced CT technique relies on inhalation of stable xenon and CBF calculation using a modified Kety–Schmidt equation. PET is regarded to be the gold standard for CBF quantification. PET measurements are performed using bolus injection of H$_2^{15}$O or continuous inhalation of C$_{15}$O$_2$, followed by CBF quantification using the Kety–Schmidt equation. The most common MRI methods for potential absolute quantification of CBF are arterial spin labelling (ASL) and dynamic susceptibility-contrast MRI (DSC–MRI). In ASL the tracer is created by inversion of spins in a brain-feeding artery. The inverted spin population is transported to the tissue with the blood and subsequently reduces the longitudinal tissue magnetisation in proportion to the regional CBF. DSC–MRI is an intravascular bolus-tracking technique based on a temporary contrast-agent induced signal loss during the bolus passage. First-pass monitoring of the intravenously injected gadolinium tracer concentration at a rate of approximately one image per second is required for subsequent calculation of cerebral blood volume, MTT and CBF. Deconvolution of the measured tissue concentration function with the arterial input function (AIF) yields the tissue residue function scaled by the CBF. Absolute quantification is at present hampered by difficulties in retrieving a reliable AIF. Finally, dynamic CT perfusion imaging is an emerging technique, based on basically the same bolus-tracking concept, using intravenous injection of an iodinated contrast material. Figure P.24 displays examples of CBF maps obtained by different modalities. The top row of Figure P.24 shows (from left to right) Tc-99m-labelled ECD SPECT (patient case), Xe-133 SPECT (normal volunteer) and H$_2^{15}$O PET (normal volunteer). The bottom row (from left to right) illustrates pulsed ASL MRI (normal volunteer), DSC-MRI (normal volunteer) and dynamic CT (patient case).

**Related Articles:** Mean transit time (MTT), Arterial spin labelling (ASL)


**Perfusion imaging**

*(Nuclear Medicine)* This refers to the imaging of the perfusion process, namely the nutritive delivery of arterial blood to the capillary bed.

In nuclear medicine perfusion studies are performed on a routine basis, e.g. in lung perfusion studies and myocardial perfusion imaging (MPI).

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**FIGURE P.24** Examples of cerebral blood flow (CBF) maps. Top row (from left to right): Tc-99m-ECD SPECT, Xe-133 SPECT, H$_2^{15}$O PET. Bottom row (from left to right): pulsed ASL MRI, DSC-MRI, dynamic CT.
In lung ventilation/perfusion studies, the examination is divided into two parts: ventilation and perfusion. The ventilation test is designed to check that air is able to reach all parts of the lung. The perfusion test is used to identify a potential pulmonary embolism. A pulmonary embolism in a vein causes the blood flow to decrease in the vein following the embolism, hence a lower relative perfusion compared to unobstructed veins.

**Abbreviation:** MPI = Myocardial perfusion imaging.

**Related Articles:** Technetium generator, Cyclotron

### Periodic motion

*Magnetic Resonance* In magnetic resonance, periodic motion refers to regular patient movement such as cardiac and respiratory motion. This type of motion can cause artefacts to occur in the images. A ghost of the image occurs in the phase encoding direction, which can cause problems if the ghost overlays a part of the image which is clinically relevant.

There are several ways in which this type of artefact can be reduced. The first is by breath-hold imaging where data is acquired in a single breath-hold. This is only useful if the pulse sequence is extremely quick and if the patient is well enough to hold their breath for the appropriate length of time. The next possible solution is gating which can be used with both cardiac and respiratory motion to ensure that the images are acquired at the same point in the cycle every time. The final possibility is to swap the frequency and phase encoding directions so that the artefact does not overlay the anatomical region of interest.

**Related Article:** Motion artefacts

### Periodic table

*General* This is a table with the 117 known basic elements organised in a systematic manner. The table is designed to demonstrate the recurring (periodic) nature of the element properties. Due to the smart design of the table, properties of an element can be concluded from its position. The elements are arranged in ascending order by atomic number. Elements with similar properties are placed in the same column. Each horizontal row represents the filling of a quantum shell in the atomic model. The rows and columns are also referred to as periods and groups respectively. Each element is represented by its specific element symbol and the atomic number.

### Peripheral blurring

*Diagnostic Radiology* See Unblanking

### Peripheral dose

*Radiation Protection* The peripheral dose is the unintentional dose which is delivered to tissue outside the primary radiation field. The common sources of this peripheral dose may be leakage radiation from the radiotherapy unit treatment head, or scattered radiation from the collimators and from interactions between the radiation beam and the air before the beam reaches the patient. Routine checks are carried out on radiotherapy units to ensure that the peripheral dose is as minimal as possible.

### Peripheral nerve stimulation

*Magnetic Resonance* A peripheral nerve stimulation (PNS) can occur in patients undergoing MRI or MRS if magnetic field gradients are switched rapidly. A high switch rate results in a short rise time for the gradient field and a subsequent rapid change of the magnetic flux density in the patient. The time for the gradient fields to reach the maximum gradient amplitude, which is expressed in mT/m, is known as the rise time, measured in units of microseconds. The faster the rise time the greater is the likelihood of PNS.

Typical gradient systems are capable of producing gradients from 20 to 100 mT/m. The PNS has been studied in volunteers under different experimental circumstances. PNS occurs above a threshold which is expressed in terms of dB/dt given in mT/s. The PNS threshold value is proportional to \( t^{0.5} \), where \( t \) is the switching time of the gradients. The modulus of the gradient vector field \( |B| \) is closely correlated to the PNS threshold level than \( B \), the imaging component of the gradient field. In normal MRI sequences the induced currents are of a few tens of milliamperes per square metre a value that is below current densities present in the normal brain and heart tissue. In single-shot techniques the rapid switching of magnetic field gradients is able to produce induced currents that exceed the nerve depolarisation threshold and cause peripheral nerve stimulation. The number of gradient switching operations per unit of time determines the duration of the induced current while the rate of change of the magnetic field determines the maximum amplitude of the induced current also referred as the maximum slew rate with a given gradient system. The likelihood of PNS occurring is greatest during echo-planar imaging, mainly in the acquisitions in oblique planes of section, since a higher slew rate results from the combined contributions of gradients from more than one axis. The likelihood of PNS is also greatest when the readout gradient at echo-planar imaging is in the craniocaudal direction.

### Permanent implant

*Radiation Therapy, Brachytherapy*

**Duration of the Treatment in Brachytherapy:** There are two different types of brachytherapy treatments. These are permanent and temporary depending in general on the time for which the radioactive material is applied:

1. **Permanent implants**
   a. The sources are placed in the target volume by an interstitial technique.
   b. The sources are then left permanently in the target to decay.
   c. The total dose is delivered over a long time with decreasing dose rate.
   d. There is just one implant procedure for the whole treatment.
   e. This is a low dose rate technique.
2. **Temporary implants**
   a. The sources are placed in the target volume by either interstitial or intracavitary techniques.
   b. The sources stay in the target volume until the correct dose is delivered.
   c. Fixation of applicators can be used for short HDR treatments, with good control of both total dose and dose distribution.
   d. The treatment is often delivered in several fractions, i.e. several implant procedures are required.
   e. Temporary implants are suitable for all dose rates, high dose rate, low dose rate and pulsed dose rate.

Permanent implants are often used today for the treatment of ‘early small prostate tumours’ (low risk patients). Small radioactive seeds are placed in the prostate according to a planned pattern under transrectal ultrasound guidance (see Figure P.25), using a template to position the interstitial needles. The most common source used is \(^{125}\text{I}\) (half-life 59.4 days, average photon energy 28.4 keV); \(^{103}\text{Pd}\) (half-life 170 days, photon energy 20.1 and 23.0 keV) is also used (Figure P.26).

**Related Articles:** Brachytherapy, Interstitial brachytherapy, Temporary implant, Iodine-125
Permanent magnet

A magnet where the magnetic field originates from a permanent ferromagnetic material is called a permanent magnet. The permanent magnet doesn’t need any power supply or cooling as the electro- or superconductor magnets. The iron-core limits the Fringe-field but the weight of the magnet limits the magnetic field, $B_0$, to 0.4 T. The weight and the cost of the magnet is the main disadvantage and there are also inhomogeneities in the permanent magnetic field, limiting the field-of-view (FOV) (Figure P.27).

Related Articles: Electro-magnet, Field-of-view (FOV), Fringe-field, Magnet, Resistive magnet, Superconductive magnets

**Permeability-surface area product**

(Nuclear Medicine) The permeability-surface area ($PS$) product is a composite constant that describes the extractive properties of the capillary bed. $PS$ is the product of permeability $P$ for the capillary wall and the surface area $S$ available for transfer and has the same unit as flow, that is mL/h. $PS$ relates to the blood flow $Q$ and the extraction fraction $E$ (see Related Articles) by

$$PS = -Q \ln(1 - E)$$

or,

$$E = 1 - e^{-PS/Q}$$


Personal protective equipment (PPE)

(Radiation Protection) Exposure to external radiation may be restricted or reduced by the use of appropriate personal protective equipment (PPE), either worn by an individual (such as lead aprons), or placed between the individual and the source of radiation (e.g. bench shields). PPE is assumed not to include structural radiation shielding such as lead-lined walls and doors, or lead-glass windows.

The use of PPE is regarded as the last resort in the hierarchy of control measures identified by a radiation risk assessment when planning a facility designed for carrying out work with ionising radiation. Such control measures, designed to reduce radiation doses to staff, patients and the public to be as low as reasonably practicable (ALARP), start with engineering control, followed by the use of written safety instructions, and finally the use of PPE. These measures are described in more detail here.

In an ideal radiation facility, exposure of persons is completely avoided by containing the radiation source within a shielded room without allowing any access by persons during exposure. The boundaries of the room have appropriate structural protection (lead, concrete, etc.) such that no radiation dose from the facility is measurable outside. Furthermore, if any person attempts unauthorised or uncontrolled access to the room during an exposure, there will be locks to prevent such access, or interlocks that automatically make the room safe to enter, either by switching of the electrical supply to x-ray generating equipment, or by withdrawing a radioactive source to a shielded container.

However, in many circumstances it is necessary for persons to be present within the room during radiation exposures – for example during medical procedures such as x-ray examinations. In these situations it is necessary to have written safety instructions (sometimes called local rules or systems of work) together with operating protocols to tell those present what they should and shouldn’t do during the work procedure to reduce their radiation exposure to a minimum.

It is only if the radiation risk assessment determines that neither engineering controls and structural shielding, nor written instructions, can prevent the possibility of persons receiving a significant radiation dose that the use of PPE should have to be considered. If it is considered necessary, then the radiation employer has a duty in law to provide appropriate and suitable PPE, and to ensure that it is maintained. Employees also have a duty in law to wear or use any such PPE that is provided.
Related Articles: Risk assessment, As low as reasonably practicable (ALARP), Lead apron


Personnel dosimetry
(Radiation Protection) Radiation workers must be monitored to ensure that the dose they receive from their occupational exposure does not exceed the dose limits. Dosimeters used for personnel dosimetry may be active or passive devices. Dosimeters have also been designed to monitor specific parts of the body.

Related Articles: Personal dosimetry, Whole body dosimeters, Extremity dosimeters, Dose limits.

Perspex characteristics
(Radiation Protection; General) Perspex is a thermoplastic – a polymer that turns to liquid when heated, but is solid at normal ambient room temperatures (melting point approximately 140°C). It is also transparent at optical wavelengths (it is transparent from approximately 300 to 2800 nm); however, it is less than half the density of normal plate glass. As such it can be used as a lightweight alternative, although it does have disadvantages such as being easily scratched. It is also known and commercialised under different names: Plexiglas, Lucite, etc. Chemically it is a poly(methyl methacrylate), sometimes referred to as PMMA.

Perspex has a range of uses in medicine (implants, orthopaedic surgery, etc.), as well as proving a useful material for use in imaging phantoms, and in radiation protection.

In radiation dosimetry, small strips of perspex are used during the gamma irradiation process. The optical density of the perspex changes with the gamma dose absorbed by the material, and a spectrophotometer can be used to measure the change and thus determine the absorbed dose.

In medical imaging (e.g. x-ray including CT, and nuclear medicine), perspex is often used as a material for the construction of imaging and dosimetry phantoms for quality control purposes. An example of a perspex dosimetry phantom for use in CT scanners is given in Figure P.28.

In radiation protection, perspex is frequently used as a shielding material against beta radiation, for example from the following radionuclides commonly used in medicine – 32Phosphorus (32P), 35Sulphur (35S), 14Carbon (14C), and 90Strontium (90Sr) and 90Yttrium (90Y). Perspex is used because its low effective atomic number (~6.5) means that the absorption of beta radiation by the perspex does not give rise to significant bremsstrahlung as would be the case with higher atomic number materials such as lead which is used for shielding from photon radiations – for example x-ray or gamma radiation.

The following is a summary of some of the characteristics of perspex that prove useful in medical physics:

- Effective atomic number: 6.48 (cf. water 7.42)
- Density: approximately 1.15–1.19 g/cm³ (water 1.0)
- Linear attenuation coefficient (at 70 keV): 0.2184 cm⁻¹ (water 0.1945)
- Mass attenuation coefficient (at 70 keV): 0.1836 g/cm² (water 0.1945)
- CT number: approximately +120 (water +0)
- Range of 32P beta particles in Perspex: 6.1 mm

Related Articles: Beta radiation, Bremsstrahlung, Acrylic


Pertechnetate
(Nuclear Medicine) A pertechnetate is a compound containing the $^{99m}$TcO$_4^-$ ion.

In the example of the technetium generator, the pertechnetate is extracted from the technetium generator using a liquid solution of sodium chloride. The extracted compound is sodium pertechnetate (Na TcO$_4$) and it is the basis for a wide diversity of radiopharmaceuticals used for diagnostic purposes.

PET (positron emission tomography)
(Nuclear Medicine) See Positron emission tomography (PET)

PET clinical applications
(Nuclear Medicine) The clinical use of positron emission tomography (PET). The most common clinical use for PET is tumour localisation using $^{18}$F-FDG which targets cells with high glucose uptake, i.e. high metabolism. PET is more sensitive and has a higher resolution compared to SPECT. One downside is that most of the positron emitters used for PET imaging have a short half life; hence, it requires production onsite or at a facility close by. Radionuclide production instrumentation is also relatively expensive. Table P.3 gives the most common radionuclides used for PET imaging and their applications.

### Table P.3

<table>
<thead>
<tr>
<th>Radionuclide</th>
<th>Half-Life</th>
<th>Molecular Imaging Applications</th>
</tr>
</thead>
<tbody>
<tr>
<td>$^{18}$F</td>
<td>1.8 h</td>
<td>Metabolic activity (tumour, inflammation, infection), receptor binding, transporter function, enzyme activity</td>
</tr>
<tr>
<td>$^{11}$C</td>
<td>20 min</td>
<td>Metabolic activity (myocardium, cardiac infarction detection), receptor binding</td>
</tr>
<tr>
<td>$^{15}$O</td>
<td>2 min</td>
<td>Metabolic activity (cognitive function)</td>
</tr>
<tr>
<td>$^{13}$N</td>
<td>10 min</td>
<td>Protein synthesis, cell proliferation (mitotic rate)</td>
</tr>
<tr>
<td>$^{124}$I</td>
<td>4 days</td>
<td>Antibody binding</td>
</tr>
</tbody>
</table>

FIGURE P.28 CT Dosimetry Phantoms made from Perspex. (Graphs courtesy of ImPACT, UK, www.impactscan.org)
**Abbreviation:** PET = Positron emission tomography.


**PET/CT**

*(Nuclear Medicine)* The combination of a PET scanner and a CT scanner in a single imaging system. The PET scanner will acquire functional information from the in vivo tracer bio-distribution which is complemented by the morphological information given by the CT. The CT data can also be used to estimate a low-noise attenuation correction for PET. With such an accurate attenuation correction the PET scanning time (15–30 min) is shorter than for a comparable stand-alone PET. Using the morphological CT information it is possible to separate specific uptake (e.g. tumour-specific) from non-specific and to more accurately localise malignancies. The CT also provides additional diagnostic information, for example, tumour size or other lesions not evident in the PET image.


**Phantom** *(Ultrasound)* A phantom is an artificial object that mimics the ultrasonic interaction in the human body. It consists of a tissue mimicking material, targets and a container (Figure P.29). The acoustic properties set to imitate human tissue characteristics are usually speed of sound, attenuation coefficient and scattering characteristics, but sometimes properties such as elasticity and thermal properties are important. A common phantom material is an aqueous gel mixed with graphite particles to produce ultrasound images with a similar speckle pattern and attenuation to human tissue. Nylon wires are often used as ‘point targets’ in the imaging plane to test axial and lateral resolution and calliper accuracy (Figure P.30).

Phantoms are used predominantly for calibration and quality assurance of ultrasound equipment, training of personnel, research and development. Phantoms are designed to allow tests of important imaging parameters such as geometrical accuracy, spatial and contrast resolution, calliper accuracy, penetration and image consistency.

**Related Articles:** Quality assurance, Spatial resolution, Contrast resolution

**Phase angle** *(Magnetic Resonance)* The acquired signal from an MR measurement is complex, i.e. it has both a real and an imaginary part. The net magnetisation vector $\mathbf{M}$ (or the net magnetic moment) of an undisturbed spin population is initially parallel to the main magnetic field $\mathbf{B}_0$. Immediately after excitation, $\mathbf{M}$ is rotated away from the direction of $\mathbf{B}_0$, which produces a vector component $M_y$ of $\mathbf{M}$ in the transverse $xy$ plane, perpendicular to $\mathbf{B}_0$ (Figure P.31).
Phase coherence

(Magnetic Resonance) The phase $\varphi$ is defined as the angle that $\mathbf{M}_x$ in the transverse plane describes with one of the axes ($x,y$), where $\mathbf{M}_x$ is the projection of the total magnetisation $\mathbf{M}$ onto this plane (Figure P.31). $\mathbf{M}_x$ is the resultant of all spins that are polarised in the xy plane.

$\mathbf{M}_x$ precesses around the z-axis with the Larmor frequency. The RF-excitation that polarises the spins in the xy-plane makes all excited spins start their precession at the same angle $\varphi$. If all spins experience exactly the same external magnetic field, they will have exactly the same frequency and will rotate with the same angular velocity, corresponding to $\omega = \gamma B_0$. The spins will thus rotate synchronously, all having the same phase angle at any given point in time. In MRI, this is called phase coherence.

Related Articles: Larmor frequency, Phase contrast MRI, Phase dispersion

Phase contrast

(Magnetic Resonance) The acquired signal from a MR measurement is complex. That is, it has both a real and imaginary part, corresponding to the vector components of the magnetisation vector in the x- and y-directions in the rotating reference system, respectively (Figure P.31).

![Image of magnetisation vector](image)

**FIGURE P.31** The magnetisation vector $\mathbf{M}$ after an excitation pulse. The angle between the y-axis and the vector component of $\mathbf{M}$ in the transverse plane ($\mathbf{M}_{xy}$) is called phase angle ($\varphi$).

Most MR images are presented as magnitude images, which is a vector addition of the real and imaginary parts of the signal:

$$M(S) = \sqrt{(\text{Re}(S))^2 + (\text{Im}(S))^2}$$

where

- $M(S)$ represents the magnitude of the acquired signal
- $\text{Re}(S)$ and $\text{Im}(S)$ are the real and imaginary part of the signal, respectively

The phase of the signal is calculated as

$$\varphi = \arctan \left( \frac{\text{Im}(S)}{\text{Re}(S)} \right)$$

Ideally, the phase signal should be zero in a normal MR image. However, due to, e.g. eddy currents and field inhomogeneities, the phase value is often not zero. Likewise, if the spin is moving during imaging, it acquires a net phase offset. Generating an image of the phase distribution gives what is known as phase contrast images. This can be utilised in mapping the velocity of moving spins in the image, as the phase is related to the velocity. For further information regarding phase-contrast, see Velocity encoding.

Related Articles: Velocity encoding, Phase contrast angiography


Phase contrast angiography

(Magnetic Resonance) MR can be used to visualise blood vessels by several methods, commonly called magnetic resonance angiography (MRA). The method described next uses the phase-contrast, or velocity encoding (Phase contrast MRA [PC-MRA]) technique. With this method, bipolar gradients are used to encode velocity in three orthogonal directions (see Velocity encoding, Phase contrast).

Two image sets are recorded, with opposite velocity encoding gradients. In these two image sets, static spins will acquire phase shifts that are equal in magnitude but opposite in sign, and a magnitude subtraction cancels this signal. This property is especially attractive when performing MRA, as good background suppression is crucial for getting diagnostic images.

Moving spins, on the other hand, will accumulate phase shifts of different magnitude as well as phase. In the magnitude subtraction, a signal will remain (Figure P.32).

A PC-MRA sequence is most often a 3D gradient echo sequence with a short TR and short TE. Velocity encoding is performed in three orientations, in order to capture motion in all directions. The data is post processed with maximum intensity projection (MIP) to get a 3D overview of the vessel tree.

PC-MRA is clinically used in brain, abdomen and the extremities. For an example, see Figure P.33.

Related Articles: Contrast-enhanced angiography, Maximum (minimum) intensity projection, Phase contrast, Time of flight (TOF), Velocity encoding (VENC)

Phase dispersion

(Magnetic Resonance) The phase $\varphi$ is defined as the angle that $\mathbf{M}_x$ in the transverse plane describes with one of the axes ($x,y$), where $\mathbf{M}_x$ is the projection of the total magnetisation $\mathbf{M}$ onto this plane (Figure P.34). $\mathbf{M}_x$ is the resultant of all spins that are polarised in the xy plane.

$\mathbf{M}_x$ precesses around the z-axis with the Larmor frequency. The RF-excitation that polarises the spins in the xy plane makes all...
Phase encoding

excited spins start their precession at the same angle $\phi$. If all spins experience exactly the same external magnetic field, they will have exactly the same frequency and will rotate with the same angular velocity, corresponding to $\omega = \gamma B_0$.

However, in a realistic environment, the external field is neither static nor homogenous. On a microscopic scale, each spin is affected by the time-varying field from its neighbours. On a macroscopic scale, the external field $B_0$ is modulated by inhomogeneities that can result either from imperfect shimming or from susceptibility effects within the body. Within each voxel, spins in different subvolumes may therefore experience different effective magnetic fields. Each subvolume will thus contain spins that rotate with an angular velocity $\omega$ corresponding to the sum of the magnetic fields that it experiences, $\omega = \gamma B_0 + \Delta B_0$.

Even if all spins start their rotation in the $xy$ plane with the same phase angle, their slightly differing angular velocities will, over time, cause them to accumulate different phases.

In MRI, this is called phase dispersion, and causes signal loss as the length of the resulting $M_{xy}$ becomes shorter.

Related Articles: Phase coherence, Phase contrast MRI, Shimming, Susceptibility

Phase encoding

Phase encoding is one of the principal methods used for spatial localisation in MRI. It is normally used in conjunction with frequency encoding to localise signal within a selected slice.

Frequency encoding allows position along a chosen axis to be encoded into the NMR signal and recovered using Fourier transformation. This yields a projection through the object, but if the same thing could be done along the perpendicular axis as well, 2D Fourier transformation could be used to recover the entire image. However, application of two perpendicular frequency encoding gradients simultaneously would not achieve this: instead, it would result in a single projection along a direction defined by the sum of the two gradients. Phase encoding is a clever means of getting around this problem, encoding positional information into signal frequency by indirect means.

To perform phase encoding, a gradient is applied for a short time during the interval between excitation and signal acquisition. The gradient is oriented perpendicular to the frequency encoding axis in the plane of the selected slice. While the gradient is on, elements of transverse magnetisation at different locations along the gradient direction precess at different frequencies. When the gradient is switched off the magnetisation returns to a common precessional frequency, but the phase dispersion accumulated in the presence of the gradient remains. This phase is given by the equation $\phi(y) = \gamma G_{y} t_y$, where $y$ is the displacement along the gradient direction, $G_y$ is the gradient amplitude (in mT m$^{-1}$), $t_y$ is the gradient duration, $\gamma$ is the gyromagnetic ratio of the nucleus (normally $1H$) and $B_0$ is the strength of the static magnetic field. Frequency encoding is then applied along the perpendicular direction as normal. The acquired echo contains positional information into signal frequency by indirect means.

This entire data acquisition process is repeated a number of times, corresponding to the desired image resolution (typically 256, 512 or, increasingly, 1024). On each repetition a different phase encoding gradient amplitude is used, so that the phase of the signal from magnetisation at a given position along the phase encoding gradient direction is incremented. The phase evolution across the set of acquired echoes mimics the effect of frequency...
Phase image

**Phase image** *(Magnetic Resonance)* A phase image is composed of the phase values, transformed into a greyscale, of the acquired MR signal.

In conventional MRI, magnitude images are used. These images represent the signal amplitude of the transverse magnetisation. However, the signal also has a phase value, which can be used for imaging as well. These maps are called phase images.

In an ideal MR experiment, the spins are encoded to have a specific phase value depending on their position in space. But because of, for example field inhomogeneities and motion, the phase value can be different. As the phase value is used in the Fourier transform for positioning data in the Fourier domain, an erroneous phase value can cause image artefacts like geometric distortions.

Phase images are used in various MR experiments. For example, in phase-contrast MR, phase images are used to depict moving spins. Phase images can also be used in mapping field inhomogeneities or in MR-venography or SWI.

**Phase quadrature** *(Magnetic Resonance)* Two signals are said to be in phase quadrature if they are +90° or −90° out of phase.

The cosinusoidal signal

\[
s(t) = A_0 \cos(\omega t)
\]

is in quadrature to the sinusoidal signal (Figure P.37):

\[
s(t) = A_0 \sin(\omega t)
\]

**Phase quadrature** *(Ultrasound)* In general, phase quadrature means 90° out of phase. Alternatively we can say that when two signals differ in phase by −90° or +90°, they are said to be in phase quadrature. An important quantity of two signals in phase quadrature is that the expected value of their product, is zero, which can be useful to check whether two signals are in phase quadrature. Signals in phase quadrature are used in quadrature detection for separating forward and reverse flow components in Doppler systems for example.

**Phased array coil** *(Magnetic Resonance)* A phased array coil is an MR receiver coil consisting of an array of individual receiver coils. A phased array yields the high signal to noise ratios seen with small surface coils while simultaneously providing a large field of view.

In a phased array adjacent coils are overlapped to eliminate mutual inductance. Mutual inductance is a measure of coupling between coils. This coupling would cause an unwanted splitting of the tuned resonant peaks of the coils and a reduction of coil sensitivity. The degree of overlap required to set the mutual inductance to zero is determined by the geometry of the individual coils.

Each coil in the phased array is connected to a low impedance pre amplifier. Use of low impedance amplifiers helps reduce coupling between distant coils. The outputs of the preamplifiers in the array are sampled simultaneously and combined electronically (Figure P.38).

A single receiver coil may then consist of many individual coil ‘elements’. Parallel imaging techniques depend on the use of multi-element array coils. Differences in spatial sensitivities of each element are exploited in parallel imaging to deliver scan time savings (see Parallel imaging).

Phased array transducer

(Ultrasound) The phased array transducer produces a sector-shaped scan by simultaneously firing all of its piezoelectric transducer elements (typically 128 or 256) for each scan line (Figure P.39). This is contrary to the linear array, which only uses parts of its total elements per line. The spacing between the elements is usually below λ/2, where λ denotes the centre frequency, to reduce grating lobe effects.

The other difference compared to linear arrays is that to produce the sector-shaped image the ultrasound beam is steered in all directions within the sector. This beam steering is accomplished by introducing individual time delays to the transducer elements in a similar way as with electronic focusing in a linear array (Figure P.40).

In normal use the operator will angle the transducer in such a way that the field of interest is in the straight-ahead direction. This is because the beam width at the focal point will widen with the steering angle, resulting in a worse lateral resolution in the outer parts of the scan sector. This can be understood by the fact that the effective aperture (i.e. the aperture seen from the point of view of the propagating beam) of the transducer gets smaller with the steering angle (Figure P.40).

The phased array transducer is generally used for cardiac imaging as the small aperture fits in between the ribs, and the large sector-shaped scanning area covers the heart.

Related Articles: Linear array, Grating lobes

Phosphocreatine

(Magnetic Resonance) Phosphocreatine is a chemical compound that features in in vivo phosphorus (31P) NMR spectra of a number of organs. In muscle and brain spectra, the PCr resonance is usually the most prominent peak in the spectrum, and may be used as an internal chemical shift reference (Figure P.41).

The importance of phosphocreatine in 31P NMR arises from its role in the body’s energy metabolism. PCr acts as a source of ATP, and is depleted to maintain ATP levels during ischaemia and hypoxia. Thus PCr levels can be used to monitor the effect of fatiguing exercise in skeletal muscle (Figure P.42). It also provides a prognostic indicator in birth asphyxia that is well correlated with neurodevelopmental outcome (Figure P.43).

Phosphodiesters (PDE)

(Magnetic Resonance) Phosphodiesters are chemical compounds in which a phosphate group is joined to two other molecules via ester bonds. These compounds are found in vivo, and feature in phosphorus (31P) NMR spectra of a number of organs.

The PDE region of an in vivo 31P spectrum occupies the chemical shift range around 3 ppm (relative to phosphocreatine). It consists of a broad peak due primarily to glycerol 3-phosphorylethanolamine and glycerol 3-phosphorylcholine. The individual compounds cannot normally be resolved (Figure P.44).

Biochemically, these compounds are catabolites of phosphatidylethanolamine and phosphatidylcholine, which are major components of membrane phospholipids and of myelin in the brain. Increases in PDE levels are often observed in malignant tumours, due to increased membrane metabolism (Figure P.45).

Phosphomonoesters (PME)

(Magnetic Resonance) Phosphomonoesters are chemical compounds in which a phosphate group is joined to another molecule via ester bonds. These compounds are found in vivo, and feature in phosphorus (31P) NMR spectra of a number of organs.

The PME region of an in vivo 31P spectrum occupies the chemical shift range between about 5 and 7.5 ppm (relative to phosphocreatine). It consists of a broad peak due primarily to phosphocholine, phosphorylethanolamine and sugar phosphates. The individual compounds cannot normally be resolved (Figure P.46).

Biochemically, these compounds are precursors of phosphatidylethanolamine and phosphatidylcholine, which are major components of membrane phospholipids and of myelin in the brain. Increases in PME levels are usually observed in malignant tumours, due to cell proliferation and associated increased membrane metabolism (Figure P.47).
A phosphor is a substance that exhibits phosphorescence, the prolonged emission of light following exposure to radiation. Phosphorescence is a type of photoluminescence, similar to fluorescence, which immediately re-emits the absorbed radiation. The study of phosphorescence contributed to the discovery of radioactivity in 1896.

Phosphors are various transition metal or rare earth compounds. Phosphors are usually made by adding an 'activator' to a 'host material'. The host materials are often oxides, sulphides or halides of zinc, manganese or aluminium. The activators sustain the emission time or 'afterglow'. Other materials, such as nickel, may be used to shorten the afterglow. In general, the persistence of the afterglow increases as the wavelength of the emitted light increases. Common phosphors include copper-activated and silver-activated zinc sulphide, which emit green light.

Phosphors are commonly used to produce 'glow-in-the-dark' objects, cathode ray tube displays and fluorescent lights. They are added to paints and cosmetic creams for 'glow-in-the-dark' face paints. Phosphors, such as zinc sulphide, were once used to paint dials of watches by mixing them with radioactive substances, such as radium, to excite the phosphor. Materials which exhibit electroluminescence produce light sources from a large area by excitation with an electric field. These substances are suitable for backlights, such as those used in a liquid crystal display (LCD).

**Medical Applications:** One of the most common phosphors is sodium iodide doped with thallium, NaI(Tl), used in scintillation detectors due to its high light output. NaI(Tl) crystals are often
coupled with photomultiplier tubes, used in gamma cameras for nuclear medicine. Control of crystal growth is used to adjust its parameters, such as radiation hardness, afterglow and transparency. NaI(Tl) is also used in x-ray detectors. Caesium iodide (CsI) is regularly used as the input phosphor of x-ray image intensifiers for fluoroscopy. ZnS is used as a scintillation detector for x-ray screens and cathode ray tubes. Lithium fluoride (LiF), known as TLD-100, is used to record gamma and neutron exposure in thermoluminescent dosimeters. Many other TL dosimeters are also known such as calcium sulphate (CaSO₄), calcium fluoride (CaF₂), lithium borate (Li₂B₄O₇), and aluminium oxide (Al₂O₃).

**Related Articles:** Gamma ray, Gamma camera, Image intensifier, Fluoroscopy, Neutron, Nuclear medicine, Radioactivity, Radium, Thermoluminescent dosimeter (TLD), X-ray

**Phosphor layer**

*(Diagnostic Radiology)* In conventional diagnostic x-ray using film, a fluorescent intensifying screen is used. The screen contains phosphors which fluoresce on exposure to x-ray radiation. The visible light emitted from the phosphors blackens the film and a latent image is formed. The phosphor layer is the layer of the screen containing phosphor particles suspended in a supporting substrate. The thickness of the phosphor layer is in the range 100–300μm. Thinner phosphor layers will support higher resolution as there is less dispersion of light in the phosphor thickness. A thicker phosphor provides a higher speed, as more x-ray interactions will occur in the thicker material.

**Phosphorescence**

*(Nuclear Medicine)* Phosphorescence is a slow process that briefly prevails beyond the exciting source being removed (afterglow), as opposed to fluorescence which is a prompt (fast) process that will generally not exist beyond the exciting source being removed. Phosphorescence occurs when electrons are trapped in an impurity-induced energy state where further de-excitation is forbidden. The electrons will eventually de-excite to the ground state after being excited to an adjacent energy state. This delay in de-excitation will cause some single-photon registrations after the original pulse has died out. This effect is also referred to as *afterglow*.

The scintillation property of a material is dependent on the energy states determined by the lattice structure. The electrons in such materials are only allowed in certain discrete bands, the upper two bands are called *valence band* (the lower of the two) and *conduction band* (the upper of the two). In a pure crystal these two bands are separated by a gap of ‘forbidden’ energies. The emitting of scintillation light is a part of a two-step process: (1) incident radiation excites electrons in the lower valence band up to the conduction band, (2) electrons release energy when they are de-excited, thus sending out a *scintillation photon*. Scintillation photons are not in an optimal energy range in terms of photomultiplier tube (PM tube) efficiency. In many inorganic scintillators, like NaI(Tl), an impurity is induced to create multiple energy states in the band gap. When electrons de-excite from one of the impurity-induced energy states to another, photons with the desired energy are emitted.

The electrons do not all follow the same de-excitation pattern, and some electrons will be caught in energy states where further de-excitation is forbidden. Further de-excitation is only possible if...
the electron is first excited (thermal excitation) to an adjacent energy state from which the electron is allowed to de-excite. Such a process will cause a delay in excitation, thus creating a slow component, i.e. phosphorescence. In NaI(Tl) some of the components have a decay time of 0.15 s and contribute some 9% of the total light yield. The main scintillation signal component in NaI(Tl) has a decay time of 230 ns. Since the time resolution of the PM tube is below 0.15 s, delayed events can easily be registered and discriminated. However for certain applications phosphorescence can cause problems, namely high count rate applications. In such applications, delayed pulses will overlap which prevents energy discrimination and therefore the delayed signals will create a background signal which ultimately leads to worse activity quantification and image quality.

**Related Articles:** NaI(Tl) detector crystal, Scintillators, Inorganic scintillators, Light yield in scintillation detectors, Bismuth germanate (BGO)


### Phosphorus

**Symbol:** P  
**Element category:** Non-metals  
**Mass number (A):** 31  
**Atomic number (Z):** 15  
**Atomic weight:** 30.974 g/mol  
**Electronic configuration:** 1s² 2s² 2p⁶ 3s² 3p³  
**Melting point:** 317.3 K  
**Boiling point:** 550 K  
**Density near room temperature:** 1823 kg/m³  

Phosphorus is highly reactive element that was discovered in 1669 by Hennig Brand and has several allotropes; white, red and black. Phosphorus is an essential component of living systems, since it is a major component of adenosine triphosphate (ATP), a molecule used to store the energy produced as cells respire. It is also present in bones and nervous tissue.

**Medical Applications:** In magnetic resonance imaging, phosphorus spectroscopy is performed to investigate muscle metabolism. Spectra show changes during and after exercise, which result from a change in the amount of ATP in the muscle. Phosphorus spectroscopy is also used to investigate changes in liver metabolism as a result of hepatitis and cirrhosis.

**Related Articles:** Beta particles, Half-life, Magnetic resonance imaging, Phosphorus-32, Phosphorus-33, Radioactive

### Phosphorus-32 [32P]

**Element:** Phosphorus  
**Isotopes:** 11 < N < 45  
**Atomic number (Z):** 15  
**Neutron number (N):** 17  
**Symbol:** 32P  
**Production:** Reactor, e.g. 31P(n,γ)32P β → 32S  
**Daughter:** 32S  
**Half-life:** 14.3 days  
**Decay mode:** β⁻ decay  
**Radiation:** β⁻ 1710.7 keV (max) 570 keV (mean)  
**Gamma energy:** None

**FIGURE P.47** 31P NMR spectrum of the human brain showing PME region.
Skin dose rate from 1 MBq: 120 μSv h⁻¹ at 30 cm (point source); 0.0054 μSv h⁻¹ at 1 m (10 mL glass vial)

Total absorption (electron range): 6–10 mm lucite

Biological half-life: bone >3 years, WB 257 days

Critical organ: red bone marrow, bone surfaces

$AL_{\text{inv}}$ (50 mSv): 10 MBq

Absorbed dose: 0.88 mGy MBq⁻¹ red bone marrow

Effective dose: 2.3 mSv MBq⁻¹ (oral) 0.8–3.2 mSv MBq⁻¹ (inhalation)

Clinical Applications: In nuclear medicine $^{32}$P is mainly used for radionuclide therapy of polycythemia vera (a rare blood disease characterised by an elevation of the immature red blood cells) and other neoplastic diseases, for example treatment of peritoneal or pleural effusions. It is frequently used in biomedical research as a tracer substitute for phosphor in, e.g. nucleic acid sequences (DNA-hybridism).

Related Article: P-32-Sodium orthophosphate


Photoc stimulation

(Magnetic Resonance Imaging) During a fMRI study, the subject undertakes a series of tasks known as a paradigm. One of the original proof-of-concept fMRI paradigms was that of photic stimulation (Kwong et al. 1992).

In the simplest type of photic stimulation paradigms, the visual stimulus is switched on and off alternately (in a block design, perhaps with 30 s blocks). In the brain areas which are responsive to the photic stimulation (such as the visual cortex),fMRI signal rises and falls as the stimulus is turned on and off (with a slight delay due to the delay in the blood flow response or haemodynamic response function). During early implementations of photic stimulation the subject wore a mask which covered their eyes and blocked out all ambient light. Mounted within the mask were a series of LEDs which flickered with adjustable frequency/intensity (Figure P.48).

Related Articles: fMRI (functional magnetic resonance imaging), Block design, Haemodynamic response function


Photo peak

(Radiation Protection) The process where a photon (x- or γ-ray) is absorbed and an electron is ejected from the atom is called the photoelectric effect. The energy of the photon $E_{\text{ph}}$ should be equal to or greater than the electron binding energy $E_{b}$. The energy $E_{\text{e}}$ of the ejected electron (photoelectron) is thus equal to

$$E_{\text{e}} = E_{\text{ph}} - E_{b}$$

The photo peak in the measured energy spectrum (Figure P.49) represents those events where the energy of the photon is fully absorbed and thus indicates the energy of the incident photons.

Detectors (e.g. scintillation detectors, proportional counters, or germanium detectors) that have output pulse amplitudes dependent on the energy of the incident radiation are used for x- and γ-radiation energy measurement.
Photo-activation therapy

(Phototherapy) Photo-activation therapy (PAT) is a novel therapeutic technique that aims to improve the effectiveness of targeted radiotherapy treatment. Increased tumour cell kill is achieved by the introduction of a high-Z atom into the DNA of the tumour cell followed by irradiation with x-rays whose energy is chosen to enhance the photoelectric effect on the particular atom. It is thought that this technique may find application in the treatment of brain tumours such as gliomas where conventional radiotherapy treatment can at best be described as palliative. In these cases, although radiotherapy enhances tumour control, the radiosensitivity of the surrounding healthy tissues does not permit the delivery of a sufficiently high dose required to treat such radio-resistant tumours. In vitro experiments on a rat glioma cell line have demonstrated that irradiation of cells treated with cisplatin, a widely used chemotherapy drug, with x-rays of energy corresponding to the platinum absorption K-edge increases local toxicity as would be expected if there was an enhanced photoelectric effect on the platinum atoms.

**Abbreviation:** PAT = Photo-activation therapy.

**Related Articles:** Radiosensitivity, Therapeutic effect


Photocathode

(Nuclear Medicine) The cathode that transforms incident scintillation photons to photoelectrons in a photomultiplier tube (PM tube). The transfer efficiency or sensitivity of the cathode is referred to as the photocathode quantum efficiency. The transfer from scintillation light to photoelectrons is a threestep process: first, the scintillation photon is absorbed and releases an electron in the process, the electron migrates to the cathode surface where it has to have enough remaining kinetic energy to escape the potential barrier.

The photocathode sensitivity or quantum efficiency (QE) is defined as

\[
\text{QE} = \frac{\text{number of photoelectrons emitted}}{\text{number of incident photons}}
\]

The maximum QE registered in most photocathodes are typically 20%–30%.

Photocathode of photomultiplier tubes

(Nuclear Medicine) A material or substance that releases electrons when it is irradiated by light photons (~400 nm). The most common use of photocathodes in nuclear medicine is in photomultiplier tubes (PM tubes). In the PM tube construction the photocathode follows directly after the glass entrance window. The back surface of the entrance window is coated with a photomissive substance. The substance typically consists of CsSb (caesium antimony) and other bi-alkali compounds. The electrons emitted from the photocathode are called photoelectrons. The number of photoelectrons produced per incident photon is referred to in terms of the quantum efficiency (QE). Typically the QE is 1–3 photoelectrons per 10 visible light photons. Quantum efficiency depends on the wavelength of the incident photons and it peaks at ~400 nm.

**Related Articles:** Photomultiplier (PM) tube, Dynode


Photoconductor

(Nuclear Medicine) A photoconductor is a device whose conductivity increases in response to electromagnetic radiation. In nuclear medicine, the photocathode of a photomultiplier (PM) tube is such...
Photoelectric absorption

A device. The photocathode emits electrons when struck by photons of visible light.

Related Article: PM Tube

Photoelectric absorption (Radiation Protection) See Photoelectric effect

Photoelectric effect (General) The photoelectric effect describes the process in which an incident photon interacts with a bound atomic electron. If the photon has sufficient energy to overcome the binding energy of the electron, it is displaced from its orbit. The photon is completely absorbed. Any energy in excess of the electron binding energy is transferred to kinetic energy of the ejected orbital electron. This results in a vacancy in the electron shell. Other electrons cascading down from outer orbital shells fill the vacancy left by this electron. During the cascade process, characteristic x-rays are emitted. The energy of these x-ray photons may also be transferred to an atomic electron, which may be ejected as an Auger electron.

The photoelectric effect dominates interaction processes at low photon energies (up to \( \sim 50 \text{ keV} \) in water or human tissues).

The interaction cross-section for the photoelectric effect varies with \( Z^3 \) and \( 1/E^3 \), where \( Z \) is the atomic number of the medium and \( E \) is the energy of the incident photon.

Related Articles: Auger electron, Characteristic x-rays

Photoelectric interaction (Radiation Protection) See Photoelectric effect

Photoelectric relay (General) An electric relay which can be controlled using light. A range of devices exists incorporating many forms of light sensitive detectors to trigger the switching of electrical contacts when a certain level of light is detected.

Photoelectric relays now normally use semiconductor light sensors with inbuilt amplifiers and comparators to drive either electromechanical relays or semiconductor switches.

They can commonly be found in situations where switching needs to be performed in high voltage environments (x-ray generators) or high insulation conditions (for patient safety), as the light may be provided from an electronic circuit via LED or incandescent bulb, and fed to the photoelectric relay through air or other insulating medium such as fibre-optic cable.

Photoelectric relays may also be used directly where it is necessary to provide an alarm or automatic action when an area becomes too dark or light for safe working.

Related Article: Relay

Photoelectron (General) An electron ejected from atomic orbit due to interaction with a photon or other incident ionising radiation. The energy of the incident photon is totally absorbed and transferred to the electron.

In a photomultiplier (PM) tube photoelectrons refer to the electrons generated by incident scintillation light in the photocathode. The photocathode is typically situated after the scintillation crystal in a PM tube. The photoelectrons (i.e. the signal) are accelerated and focused on a dynode where the incident photoelectron produces several secondary photoelectrons which are then focused onto another dynode. This process is repeated until the signal is considered strong enough to measure.

The number of photoelectrons produced per energy of incident scintillation photons is called the photocathode quantum efficiency. The maximum quantum efficiency of a photocathode is typically 20%–30%.

Related Articles: Photoelectric effect, Photocathode

Photographic detail (Diagnostic Radiology) See Detail resolution

Photomultiplier (PM) tubes (Nuclear Medicine) A signal amplifier used in nuclear medicine to magnify the signal from the scintillator. When ionising radiation interacts within a scintillator a number of light photons are created. When coupling the scintillator crystal to a photomultiplier tube (from here on abbreviated as PM tube(s)) the signal is magnified up to \( 10^6 \) times depending on PM tube design.

Using a NaI (Tl) scintillator an average of 38 light photons is created for each keV deposited in the scintillator. If these photons were to be directly transferred to an electric signal (i.e. using a photocathode) the signal would be almost undetectable among all background noise. PM tubes can produce a detectable electrical signal from even such a weak light signal.

The electric design of a PM tube is displayed in Figure P.50.

FIGURE P.50  PM tube connected to amplifier, pulse height analyser and scaler as part of a scintillation detector.
A photon is a quantum of electromagnetic radiation. It may be thought of as a ‘packet of energy’ or as a ‘particle’. Photons have no charge and, in a vacuum, travel at $3 \times 10^8$ m/s (the speed of light).

Electromagnetic radiation can be described in terms of waves or as photons – the two approaches are entirely complementary. The wave approach is most applicable at the lower energy end of the electromagnetic spectrum where one is normally dealing with large numbers of photons – a typical electric light bulb emits about $10^{20}$ visible light photons per second – whilst a nuclear medicine test may involve a million photons of gamma radiation being emitted per second from the radioactive material used and each photon can be counted separately.

Plank’s constant ($h$) links frequency ($v$) and energy: $E = hv$, and the relationship between energy and frequency is given by: $v = c/\lambda$, where $c$ is the velocity of light; hence the relationship between energy and frequency is $e = h\nu\lambda$.

The energy of photons is widely described in terms of the electron volt (eV), although the standard unit for measuring energy is the Joule (J): One eV is approximately $1.6 \times 10^{-19}$ J (Figure P.52).

Related Articles: Gamma radiation, Photons, interaction in matter

**Photon beam**

(Radiotherapy) Photon beams used in external beam radiotherapy can be produced as gamma radiation emitted by a radioactive source such as $^{60}$Co or as x-ray radiation created by accelerated electron hitting a target (bremsstrahlung radiation). The $^{60}$Co source emits two gamma rays of 1.17 and 1.33 MeV but mean photon energy of 0.9 MeV is generally accepted for the clinical $^{60}$Co beam because of the presence in the beam of low energy photons resulting from Compton interaction of the primary photons with the collimating system. The bremsstrahlung x-ray beams present a wide continuum energy spectrum whose maximum energy is approximately equal to the energy of the electron beam and the mean energy is between 1/3 and 1/2 of the maximum energy. Kilovoltage x-ray beams formed at an accelerating potential ranging from <50 kV to 300 kV have been used in radiotherapy for many years but nowadays photon beams produced by linear accelerators have superseded the kilovoltage x-ray beams in many applications. In kilovoltage x-ray beams the description of the beam quality is given by the beam half-value layer (HVL) of aluminium or copper while the specification of a photon beam produced by a linac is indicated by the value in megavolt (MeV) unit of the nominal accelerating potential of electrons that produce the photon beam. In this case the linac beam quality is given by the tissue phantom ratio (TPR). Linear accelerators

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**Figure P.51** PM tubes arranged over the scintillating crystal of a gamma camera.

Low energy photons from interactions in the crystal are lead to the entrance window. The entrance window is coated with photoemissive substance. This substance is called the photocathode and it will eject electrons when radiated by low energy photons. The efficiency (number of electrons per incident photon) is referred to as quantum efficiency. The electrons emitted from the photo-emissive substance are referred to as photoelectrons. The quantum efficiency is typically 0.1–0.3 photoelectrons per photon. The quantum efficiency is dependent on wavelength of the incident photons and for most photocathodes it peaks at $\sim 400$ nm which is visible blue light.

The photoelectrons are released into a long tube were they are accelerated by a potential difference between the photocathode and a metal plate called a dynode. The photoelectrons are directed by a focusing grid so that they strike the dynode. When the dynode is hit by the photoelectron a number of new electrons are released, these are called secondary electrons. The secondary electrons are then accelerated by a potential difference between the first dynode and a second one where new electrons are released. This procedure is repeated typically 9–12 times and the typical electron multiplication factor is 3–6 (depends on the potential difference between the two dynodes). If the PM tube contains 10 dynodes and the multiplication factor is 6, the total number of electrons registered at the end of the PM tube is $6^{10}$ ($6 \times 10^9$).

As a result, the PM tube has enhanced an undetected signal into a larger detectable signal.

In PET, and some cases SPECT, the PM tubes are placed in a ring around the patient. In most SPECT and gamma cameras the PM tubes are placed in a square, creating a detector head. In SPECT two or more of these detector heads are rotated around the patient to attain projections from all angles (Figure P.51).

**Abbreviations:** PM tubes = Photomultiplier tubes and SPECT = Single photon emission computed tomography.

**Related Articles:** PET, SPECT


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are used to produce photon beams usually in the 4–25 MeV range. Advantages of increasing the energy of the bremsstrahlung radiation consist of a deeper penetration in patients and also of the reduction of the relative (differential) absorption of photons in various tissues such as fat, bone or muscle as the Compton effect becomes the main interaction process. Other advantages are in patient skin sparing, small penumbra and reduced side scatter.

Related Articles: Tissue phantom ratio (TPR), Half value layer (HVL)

Photon fluence

(Radiation Protection) Photon fluence ($\Phi$) is a measure of the radiation field. If we consider a sphere of cross section, $a$, then

$$\Phi = \frac{dN}{da}$$

where $N$ is the number of particles incident on the sphere.

Photon flux

(Radiation Protection) Flux ($\Phi$) is defined as the average number of photons ($N$) per unit time ($t$) that pass through a unit area ($A$) perpendicular to the direction of propagation of energy:

$$\phi = \frac{N}{At}$$

Photon scattering

(Radiation Protection) Photons are scattered by interactions with orbital electrons of absorber atoms. During scattering the photon loses energy and momentum to the electron(s) it interacts with and moves off at an angle. As energy is lost in the process, there is a change in the frequency of the photon according to the formula Energy ($E$) = Planck constant ($h$) x Frequency ($\nu$).

Planck constant, $h = 6.626 \times 10^{-34}$ J s

Several different types of photon scattering are described:

Compton scattering
Rayleigh scattering
Thomson scattering

These interactions are described in more detail under the corresponding headings.

Related Articles: Compton scattering, Rayleigh scattering, Thomson scattering

Photon(s): annihilation in positron decay

(General) Positron decay is a radioactive process in which a proton in the nucleus changes into a neutron and a positively charged electron (positron or beta plus particle, $\beta^+$) is emitted, together with a neutrino ($\nu$): $p^+ \rightarrow n + \beta^+ + \nu$. Gamma rays may also be emitted.

After emission, the positron ($\beta^+$-particle) is slowed (looses kinetic energy) via interactions with surrounding matter. When brought to a near or total halt, the positron forms a particle, a so-called positronium (Ps), with an electron. The positronium (with lifetime measured in picoseconds) is consumed in an annihilation process that emits two photons with a back to back direction:

$$\beta^+ + e^- \rightarrow Ps \rightarrow \gamma + \gamma.$$

The annihilation is an example of the conversion of mass into energy and the energy of the two photons is given by applying Einstein’s equation ($E = mc^2$) with the consequence that each photon has an energy of 0.51 MeV. However, the particles are seldom brought to a full stop and the residual momentum results in a small deviation from a 180° emission angle.

Related Articles: Beta decay, Beta+ radiation, Electron capture, Photons, Positron decay, Radioactive decay

Photon(s): interaction in matter

(Radiation Protection) There are several different ways in which photons may interact with matter in the diagnostic energy range. These processes are described in detail in separate articles:

- Photoelectric effect
- Compton scatter
- Rayleigh scatter
- Pair production
- Photodisintegration
- Thomson scatter

These interactions comprise absorption and scattering of photons from an incident photon beam. The sum of these processes is equal to the total attenuation of a beam of photons passing through matter.

Related Articles: Photoelectric effect, Compton scatter, Rayleigh scatter, Pair production, Photo disintegration, Thomson scatter, Attenuation

Photon(s): mean free path

(Radiation Protection) This term is described in the article Mean free path.

Related Article: Mean free path

Photon(s): secondary

Secondary photons are those that have been removed from the primary photon beam as a result of scattering interactions.

Related Articles: Scattered radiation, Secondary radiation, Secondary ionising radiation

Photosensitivity

(Radiation Protection) The sensitivity of skin to the effects of optical and ultraviolet (UV) radiation. Exposure of skin to optical/UV radiation will invoke erythemas of varying severity dependent on radiation dose and the individual’s photosensitivity. Abnormalities of photo-sensitivity (extreme reactions to exposure) are broadly categorised into idiopathic (presumed immunological origin), genetically inherited and biochemical aetiologies. The trigger in each of a variety of dermatological conditions is either optical or UV radiation.

Overly photo-sensitive conditions have an abnormally low deterministic threshold for effect. Furthermore, normal humans of different skin-types exhibit different photo-sensitivities for the severity of erythema. Caucasians are generally more photo-sensitive than Afro-Caribbeans, for example.


Photostimulable phosphor plate

(Diagnostic Radiology) See Storage phosphor

Related Articles: Computed radiography, Storage phosphor, Fluorescence, Thermoluminescence
Photostimulated luminescence

(Diagnostic Radiology) This is a process by which energy is trapped in a phosphor and then released from the phosphor by the use of a laser. It is the basis of computed radiography (Figure P.53).

The drawing shows the basis of a storage phosphor. X-ray energy causes the promotion of electrons which are captured in f-trapping centres. A laser is used to release the energy which is captured to form an image. See the article on Storage phosphor.

Related Articles: Computed radiography, Storage phosphor

Physical penumbra

(Radiotherapy) See Penumbra

Physical phantoms

(Nuclear Medicine) Physical phantoms are used both to simulate the conditions being investigated during a patient examination and to measure the emission camera parameters (spatial resolution, sensitivity, linearity and uniformity). Phantoms used for flood measurements are one example of phantom use in a patient situation.

There are a number of phantoms used to measure the imaging parameters in an emission camera system. For example, a common set-up to measure spatial resolution is to use a bar phantom. The bar phantom consists of a number of line sources with a decreasing distance between the line source when moving along the bar phantom. A subjective approach to estimate the spatial resolution is to estimate where one can see the last visible line pair, i.e. before the lines overlap. A more quantitative approach is to use a point-spread function or a line-spread function where one uses a point and a line source phantom respectively.

Related Article: Software phantoms


Picture archiving and communication system (PACS)

(Nuclear Medicine) A computer and network system dedicated to storage, retrieval, distribution and presentation of medical images is called a PACS system. The most common storage format in hospital PACS is the DICOM format. Some advantages with a PACS system compared to imaging with conventional film are remote access for physicians and a more convenient storage process. When physicians access a PACS system from an off-site location it is commonly referred to as tele-radiology.

Related Article: DICOM (digital imaging and communication in medicine)

Piezoelectric crystal

(Ultrasound) The piezoelectric crystal is the active actuator and sensor in a conventional ultrasonic transducer. It converts electrical energy to acoustical and acoustical energy to electrical. The most common material used is PZT–lead zirconate titanate.

Related Articles: Transducer, Backing material, Matching layer, PZT, Lens, Linear array

Pincushion distortion

(Diagnostic Radiology) Pincushion distortion is an image aberration where an image appears ‘pinched’ towards its centre. It is mainly associated with image intensifiers, where it is caused by a non-uniform magnification across the image, with magnification increasing away from the image centre (Figure P.54).

Pincushion distortion

(Nuclear Medicine) The pincushion image distortion is an effect of camera non-linearity. Non-linearity effects occur when the signal in the photomultiplier tubes (PM tubes) does not change linearly with the displacement of a source across the detector face. Consider a source that moves from the edge of a PM-tube to the centre. If the light collection efficiency increases more rapidly rather than linearly as the source approaches the PM tube centre, the x and y-position signals will also change in a non-linear way, thus creating the characteristic pincushion distortion. The typical pincushion distortion of a straight-line object is apparent in Figure P.55. The opposite distortion is called a barrel distortion and is discussed in a separate article.

Related Articles: Photomultiplier (PM) tubes, Barrel distortion

![Figure P.53 Principle of photostimulated luminescence. (Courtesy of J.A. Seibert.)](image-url)
Pinhole collimator

(Nuclear Medicine) A collimator design which typically consists of a metal cone made from high absorbing material and a pinhole insert. The main clinical use of a pinhole collimator is to attain magnified images of small organs like the thyroid. Pinhole collimators play an important role in pre-clinical studies with small animal SPECT. The collimator insert is placed at the end of the high absorbing metal cone. The length of the metal cone is typically 20–25 cm.

Gamma rays that pass through the collimator will project an inverted image on the detector. The projected image will be: magnified if the distance $b$ between the collimator and the object is smaller than the collimator to detector distance $f$; and: minified if the conditions are the opposite. The relationship between image size $I$, and object size $O$ is

$$\frac{I}{O} = \frac{f}{b} \quad (P.21)$$

When the distance between the object and collimator changes, so does the imaged area. If the object is further away from the collimator a larger object can be imaged with a lower magnification; if the object is closer to the collimator a small object can be magnified to a greater extent. The relationship between the detector diameter $D$ and the image area projected onto the detector $D'$ is determined by the magnification factor $l/O$:

$$D' = \frac{D}{l/O} \quad (P.22)$$

From Equation P.22 it follows that a large magnification factor for an image acquired from an object close to the collimator results in a small imaged area.

There are some limitations when imaging 3D objects since the magnification depends on the source to collimator distance. Differences in object depth, i.e. different magnifications may result in image distortions and the user must be well aware of this effect or else it may ultimately lead to misdiagnosis and/or ineffective/harmful treatment.

The pinhole collimators contribution to the system resolution is

$$R_{\text{coll}} = d_{\text{eff}} \frac{(l + b)}{l} \quad (P.23)$$

where $d_{\text{eff}}$ is the effective pinhole diameter, accounting for photon penetration at the edges of the pinhole opening. A pinhole collimator provides very high resolution but also a limited field of view. A way to get a larger field of view is to move the object further away from the collimator, but this leads to a loss in efficiency. The efficiency of a pinhole collimator is

$$g = \frac{d_{\text{eff}} \cos^3 \theta}{16b^2} \quad (P.24)$$

where $\theta$ is the angle between the pinhole, the central line and the off-centre source. The efficiency will drop quickly with distance as seen in Equation P.24. Pinhole collimators are therefore typically used to image small organs like the thyroid and heart.

The effective pinhole diameter is

$$d_{\text{eff}} = \sqrt{d \left( d + 2\mu^{-1} \tan \left( \frac{\alpha}{2} \right) \right)} \quad (P.25)$$

where

- $\alpha$ is the angle between the walls of the metal cone
- $\mu$ is the linear attenuation coefficient of the cone material

Another pinhole application is the multi-pinhole application with several hexagonal holes in the pinhole device. This application is not often used in clinical imaging but rather for small animal imaging and research.

**Related Articles:** SPECT Collimator, Parallel-hole collimator, Diverging collimator, Converging collimator, Collimator, Collimator design, Collimator parameters


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Piston

(Ultrasound) The analysis of ultrasound fields is based on the assumption that a simple circular transducer element behaves as a piston.

**Related Articles:** Diffraction, Near zone, Far zone, Transducer
Pitch
(Diagnostic Radiology) See Helical pitch

Pixel
(Nuclear Medicine) The smallest possible element in an image is referred to as a pixel. Images typically consist of a 2D array of pixels. In nuclear medicine imaging each pixel contains a value proportional to the intensity of the incoming radiation to the corresponding detector element. The number of colours (or greyscale) that each pixel can represent depends on the number of bits per pixel, bpp. A pixel with 1bpp can have two different colours ($2^1 = 2$) and such images are called monochrome pictures. For each extra bit per pixel the number of colours available is doubled ($2^2 = 4$, $3$ bit: $2^3 = 8$, … $2^8 = 256$). In some systems as much as 24bpp is used (~16.7 million colours per pixel) with 8 pixels for green, red and blue.

Plain film radiography
(Diagnostic Radiology) Plain film radiography is another term used for normal x-ray radiography (producing a static x-ray image over film).

Planar imaging
(Nuclear Medicine) Images acquired from a specific projection angle are referred to as planar images. Planar images produce a 2D representation of the tracer distribution and planar imaging has played an important part in the history of nuclear medicine. Bone scintigraphy can be used to evaluate the skeletal system and to locate fractures invisible to x-ray examinations. Today, images are acquired from a number of different projection angles and compiled to a 3D tracer distribution using image reconstruction programs on computers. This technique is called tomographic imaging.

Related Article: Tomographic imaging

Planning target volume (PTV)
(Radiotherapy) The planning target volume (PTV) is used in radiotherapy in treatment planning. It consists of the clinical target volume (CTV) plus margins to account for geometrical variations and inaccuracies to ensure that the prescribed dose is actually absorbed in the CTV. It is purely a geometrical concept, and may sometimes even lie outside the patient’s body. For a given CTV, the PTV may vary with different beam arrangements.

The use of PTV as a planning volume was proposed by the ICRU in Report 50 (with addendum ICRU Report 62). This report provides a common framework on prescribing, recording and reporting therapies, with the aim to improve the consistency and inter-site comparability. It details the minimum set of data required to be able to adequately assess treatments without having to return to the original centre for extra information.

ICRU 62 added two new concepts – the internal margin (IM) and the set-up margin (SM), which together form the margin between the CTV and the PTV. The IM allows for error from physiological variation of the size and shape of the volume, and the SM allows for patient positioning and variation in the alignment of the treatment beams. The margins may vary individually for the different directions from the CTV (Figure P.56).

Abbreviation: PTV = Planning target volume.

Related Articles: ICRU, Clinical target volume (CTV), Gross tumour volume (GTV), Treated volume, Irradiated volume, Internal margin, Setup margin


Plastic
(General)

| Molar mass | $>10$ kg mol$^{-1}$ |
| Density at STP | varies |
| Melting point | varies |
| Boiling point | varies |

Plastic is a general term for a range of synthetic organic solid materials. Plastics are normally polymers with a high molecular weight, consisting of chains of several thousand repeating units of monomers of mostly carbon and hydrogen, commonly with oxygen, nitrogen, chlorine or sulphur in the backbone. The properties of plastics are varied by altering the monomer and various molecular groups. The adjective 'plastic' refers to materials which experience a permanent change of shape known as a 'plastic deformation' when strained beyond a critical amount. Materials described as plastic in this way, such as aluminium, are not necessarily classified as plastics and, conversely, not all plastics exhibit plastic deformation (Figure P.57).

Plastics can be classified in several ways, including their chemical structure (molecular units), such as acrylics and polyesters, or the chemical process of their synthesis, such as condensation and polyaddition. Other classifications or gradings are based on their properties such as thermoplastic/thermoset, biodegradability, density or tensile strength. The molecular structure of plastics may be partially crystalline, meaning that they often have a melting point as well as a glass transition. Semi-crystalline plastics include polyethylene, poly (vinyl chloride) and polyesters. Amorphous plastics include polystyrene, poly (methyl methacrylate) and thermosets.

![FIGURE P.56 Definition of target volumes as in ICRU 50.](image)

![FIGURE P.57 Chemical structures of common plastics: (a) polyethylene, (b) polypropylene and (c) polyvinylchloride (PVC), respectively from left to right.](image)
Plastics are extremely diverse in their uses due to their versatility, ease of manufacture, relatively low cost and imperviousness to water. Their uses are limited by their density, hardness and ability to resist heat, organic solvents and ionising radiation. Plastics generally have low toxicity due to their insolubility in water and chemical iner tness. However, they may contain toxic additives, such as plasticisers to make them pliable for products such as food packaging and tubing. Plastics are still too expensive for large scale uses such as buildings and bridges.

**Medical Applications:** The medical uses of plastics vary substantially from syringes to prosthetic limbs to medical equipment casing. Plastics have a range of applications in medical physics, including their use as phantom materials and tissue substitutes. They are suitable for such uses due to their low toxicity, chemical iner tness, ease of sterilisation and relatively low cost.

**Related Article:** Acrylic

### Plates, deflection

(General) See Deflection plates in cathode ray tubes

### Platinum (General)

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Pt</th>
</tr>
</thead>
<tbody>
<tr>
<td>Element category</td>
<td>Transition metal</td>
</tr>
<tr>
<td>Mass number A</td>
<td>195</td>
</tr>
<tr>
<td>Atomic number Z</td>
<td>78</td>
</tr>
<tr>
<td>Atomic weight</td>
<td>195.084 kg/kg-atom</td>
</tr>
<tr>
<td>Electronic Configuration</td>
<td>1s² 2s² 2p⁶ 3s² 3p⁶ 3d¹⁰ 4s² 4p⁶ 4d¹⁰ 5s² 5p⁶ 5d⁷ 6s¹</td>
</tr>
<tr>
<td>Melting point</td>
<td>2041 K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>4098 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>21460 kg/m³</td>
</tr>
</tbody>
</table>

**History:** Platinum was first known to pre-Columbian native Americans who extracted this naturally occurring metal from alluvial sands and alloyed it with gold to produce a white gold. In the 1740s Europeans rediscovered the element by examining artefacts from the earlier civilisation, resulting in a description of the metal being presented by William Brownrigg to a meeting of the Royal Society in 1750.

**Medical Applications:** Needle electrodes – Platinum is often used as a conducting component for needle electrodes. Needle electrodes pass through the skin and record potentials from small areas (such as motor units within muscles). Platinum is a good choice for this purpose as it tolerates sterilisation very well and tissue reactance to platinum is small.

Endovascular coiling – as a brain aneurysm treatment, tiny platinum coils can be threaded through a catheter and deployed into the aneurysm, blocking blood flow and preventing rupture. The coils are made of Platinum so that they are visible on x-ray scans and are flexible enough to conform to the aneurysm shape.

**Related Articles:** Electrode, Potential difference, Muscle, Aneurysm

### Plumbicon tube

(Diagnostic Radiology) The Plumbicon TV camera tube has been developed by Philips. The camera has a light sensitive layer (target) of Lead oxide (PbO). The operation of the camera is explained in the article Video camera tube. According to the light intensity at the target of the camera an electrical charge pattern is formed over the light sensitive layer. This light sensitive layer (usually with 1 in. diameter) is sometimes called retina. When the target is scanned by an electron beam the variously charged micro areas are discharged, respectively the varying discharging current (proportional to the charge of the layer, hence to the intensity of the incoming light) forms the video signal.

The change of the discharge current is related to change of the conductivity of the micro areas on the target. This means that changing the voltage of the layer will not alter the signal. This way the Plumbicon tubes have fixed sensitivity (fixed gain). As this camera can not apply Automatic gain, it is usually used in conjunction with an Automatic Brightness system.

Plumbicon TV tubes have small dark current and are less inert than Vidicon tubes (no ghosting effect, their time constant is 2–4 times smaller than Vidicon). This has two effects. From one side the camera is fast and can be used to image rapid movement (i.e. heart movement), as it will not blur the image. From other point of view this small inertia does not allow for the integration (averaging) of the image, what leads to increased noise in the image.

The Plumbicon camera has a characteristic curve with gamma close to 1.0. This means that the camera will have a smaller contrast range (latitude) than Vidicon, but the picture will be with higher contrast than TV tubes with gamma 0.7 (as is Vidicon). Plumbicon is more expensive than Vidicon. Newer lead oxide TV tubes include Newvicon and Saticon.

Plumbicon is better suited for fluoroscopic examination of dynamic objects (for example heart) and for cine/video/DVD record of the image.

CCD type cameras and flat panel detectors gradually replace the fluoroscopic x-ray systems with TV camera tubes.

**Related Articles:** Video camera tube, Superorthicon, Vidicon


**Hyperlink:** The cathode ray tube site. [http://members.chello.nl/~h.dijkstra19/page4.html](http://members.chello.nl/~h.dijkstra19/page4.html)

### PMMA (perspex, plexiglass, lucite)

(Radiation Protection) See Perspex characteristics

### Pocket dosimeter

(Radiation Protection) The original monitors developed for whole body monitoring were small films placed in a holder which provided different amounts of attenuation in different regions of the film to facilitate identification of the type and energy of the radiation which caused the exposure.

Thermoluminescent materials are also commonly used for whole body monitoring although these materials are less reliable at determining the source of the radiation exposure.

More recently, one personnel dosimetry company has introduced optically stimulated materials. These have much greater sensitivity than either film or thermoluminescent (TLD) materials.

In situations where there is significant risk of exposure or where exposures are expected to be high, active (direct reading) dosimeters are used. The original active devices were quartz-fibre electrometers. These were notoriously unreliable and have now been replaced by electronic devices using solid-state detectors. There are a variety of such solid state devices, commonly referred to as ‘pocket dosimeters’. In addition to displaying the cumulative dose at any instant they also have the ability to set alarms to warn the wearer when a particular exposure has been reached (Figures P.58 and P.59).
Point resolved surface coil spectroscopy (PRESS)

(Magnetic Resonance) PRESS is one of the most common spatial localisation techniques used for single voxel spectroscopy (SVS). Because it involves acquisition of a spin echo signal some time after excitation, it is particularly suited to nuclear species with long $T_2$ relaxation times, such as hydrogen ($^1$H) nuclei (protons).

PRESS is a 'single shot' technique, in that it requires a single acquisition of the pulse sequence to achieve localisation, and is not dependent on post-acquisition signal combination (as is the case with, for example ISIS).

The PRESS pulse sequence uses a selective 90° pulse followed by two selective 180° pulses. Each pulse is applied with a gradient along a different Cartesian axis, and hence they select orthogonal planes of spins. The first pulse and gradient combination is used to excite a plane of spins, and the second refocuses spins at the intersection of the two selected slices to form a spin echo. The third pulse refocuses magnetisation within the volume of interest at the intersection of all three slices for a second time, and the resulting volume-selected spin echo is acquired. The combination of pulses also generates a variety of unwanted signals, which are dephased using spoiler gradients (hatched in Figure P.60).

Use of a spin echo, rather than a stimulated echo as in STEAM, results in a factor of two signal advantage at a given echo time. However, VOI definition can be poorer, particularly at short echo times. Nevertheless, contamination with extraneous signal in PRESS is generally minimal.

Related Articles: ISIS, Magnetic resonance spectroscopy, Single voxel spectroscopy, STEAM

Point source calculation

(Radiotherapy, Brachytherapy) Factors that affect the dose distribution in a medium around a brachytherapy source are

- Distance – the inverse square law
- Attenuation – in the source itself and in the encapsulation (not applicable for an ideal point source)
- Attenuation – in the surrounding medium
- Build-up of scattered photons

For a point source, the dose distribution is spherically symmetric. For an ideal and not encapsulated point source of activity $A$, in vacuum, the air kerma rate in a point at distance $d$ is given in Equation P.26 (the usual symbols are used for the ratio of the mean mass energy transfer and energy absorption coefficients in medium and air):

$$\dot{K}_{\text{air}} = \frac{A \times \Gamma_{\text{a}}}{d^2}$$

The reference air kerma rate is thus

$$\dot{K}_R = \frac{A \times \Gamma_{\text{a}}}{d^2}$$

The kerma rate in vacuum at distance $d$ to a medium $m$ is

$$\dot{K}_m = \frac{A \times \Gamma_{\text{a}}}{d^2} \left[ \frac{\bar{\rho}_m}{\rho} \right]_m$$

The dose rate in vacuum at this point is (assuming electronic equilibrium and no losses due to bremsstrahlung)

$$\dot{D}_m = \frac{A \times \Gamma_{\text{a}}}{d^2} \left[ \frac{\bar{\rho}_m}{\rho} \right]_m = \dot{K}_R \left[ \frac{\bar{\rho}_m}{\rho} \right]_m \frac{1}{d^2}$$
Attenuation of the primary photons and the build-up of secondary photons in the surrounding medium can be described by a distance dependent conversion factor/function $\varphi(d)$. This function describes the deviation of the radial dose distribution from the inverse square law:

$$D_m = K_0 \left( \frac{\mu_m}{\rho} \right)^n \cdot \frac{I}{d^2} \cdot \varphi(d) \quad (P.27)$$

For brachytherapy sources with ‘higher’ energies, radium, caesium, iridium, etc., the dose distribution is dominated by the inverse square law, and the loss of primary photons is to a first approximation compensated for by the build-up of scattered photons. The conversion factor varies slowly with distance, and several approximations have been used based on polynomials, for example the Meisberger polynomial, and exponential functions.

For brachytherapy sources with lower energies, for example iodine and palladium used for permanent prostate seed implants, the photoelectric effect plays as more important role in the interactions between the photons and the medium. The effect is that the radial dose function decreases more rapidly with distance than for the higher energy sources.

Actual cylindrical sources can also be characterised starting from a point source type formalism, if an anisotropy function is added as a multiplicative factor to Equation P.27.

**Related Articles:** Treatment planning systems (brachytherapy), Source models, Meisberger polynomial


**Point spread function**

*(General)* Pixel or voxel values are usually linear combinations of the input. In shift-invariant systems, what sort of linear combination does not depend on where pixels are: if the value of the output at position $x$ is say some multiple of the input at $x$ plus some factors times its neighbours, then the same factors are true for an output at another position. The system is said to be shift-invariant. This implies that the mathematical effect of the imaging system can be summarised very simply, by stating how input pixel values contribute to an output value. The factor for all input pixels forms the *point-spread function* (PSF) of the system. The operation producing the output from the input is called a convolution:

$$\text{Output} = \text{Convolution of input with PSF}$$

A fundamental fact of signal theory play a role here: convolution corresponds to pointwise multiplication in the frequency domain. The shift-invariance thus translates to pointwise dependence in the frequency domain.

An ideal system would just transmit the image without modifying it: this means that an output value at $x$ is just a copy of the input value at $x$, and does not depend on neighbouring values: the PSF is 1 (or infinity for continuous systems) at $x$, 0 everywhere else. In practice, imperfect systems create a dependence on some neighbours. The distance from $x$ at which the PSF is non-zero explains how the input is spread over the output.

This figure shows the *magnitude* of the PSF for the radial sampling pattern:

- **A good PSF should have a strong peak:** this is how much of the value at the input point goes into the output.
- **The radial patterns explain long distance artefacts in radially sampled images:** We would try to 'push' them as far away as possible.
- **These ripples will cause another type of ringing artefacts in images.**

Note that if we know the PSF, algorithms exist for deconvolution which may improve the image quality, for example sharpen it. On the other hand, imperfect optics, as happened to the original Hubble space telescope, lead to space-variant point-spread functions.

**Abbreviation:** PSF = Point-spread function.

**Related Articles:** Modulation transfer function, Optical transfer function

**Point spread function (PSF)**

*(Ultrasound)* The point spread function (PSF) of an ultrasound imaging system is the image produced of a point scatterer (spatial impulse) at a given location. It is the convolution of the transmit and receive beam responses at that location together with the electro-acoustic conversion impulse responses. The PSF will vary throughout the image depending on the beam shapes, but given the knowledge of this distribution, the image can be predicted for any source distribution, provided that the system is linear.

**Poiseuille’s law**

*(Ultrasound)* The Poiseuille equation describes the relationship between pressure, flow and the internal dimensions for flow of a
Newtonian fluid in a cylindrical tube of constant diameter with steady laminar flow. Poiseuille performed his experiments using glass capillary tubes and described how the pressure drop \((P1 - P2)\) along the tube was directly proportional to the length (\(L\)) of the tube, the flow (\(Q\)) through the tube and a constant (\(K\)), and inversely proportional to the fourth power of the diameter. \(K\) varied with temperature and was later shown to be due to viscosity. This empirically derived relationship was solved by others to give the familiar form of Poiseuille’s equation where \(\mu\) is the viscosity of the fluid and \(R\) the radius of the tube:

\[
Q = \frac{\pi R^4 (P1 - P2)}{8 \mu L}
\]

The law is named after Jean Louis Marie Poiseuille, French physician and psychologist (1797–1869).

**Related Article:** Laminar flow

**Poisson distribution**

*(Nuclear Medicine)* The Poisson distribution is a discrete distribution that describes the probability of events occurring at a fixed time interval and with a known count-rate. The events are independent of each other. For small mean values the shape of the distribution is not symmetrical but instead skewed. As the mean value increases, the shape will be more symmetrical. A property of the Poisson distribution is that the variance is equal to the mean value. The Poisson distribution can be described as

\[
f(k; \lambda) = \frac{\lambda^k e^{-\lambda}}{k!}
\]

where

- \(\lambda\) is the expected number of occurrences during a time interval
- \(k\) is the number of occurrences

The Poisson distribution can be applied in cases where there are a large number of possible events but where each event has a very low probability. An example of this is radioactive decay where there are usually a large number of atoms but relatively few decays within a time interval. The photon fluence that strikes a scintillation camera is also Poisson distributed since the photons originated from radioactive decays. This means that when performing ROI analysis on an image, the standard deviation of a count measurement is equal to the square root of the mean count level in the ROI.

**Poisson noise**

*(Nuclear Medicine)* In scintillation camera imaging the photon fluence and thereby the collected events are Poisson distributed. Poisson noise is then the effect of this Poisson distributed fluence. When modelling an imaging system by, e.g. Monte Carlo methods it is common to add Poisson noise to the simulated images to obtain realistic noise levels comparable to clinical situations. The Poisson distribution will then follow

\[
p(x) = \frac{\mu^x e^{-\mu}}{x!}
\]

where \(\mu\) is the expectation value.

**Related Articles:** Poisson distribution, Monte Carlo

**Polar coordinates**

*(General)* The polar coordinate system is a 2D reference system where each point is associated with an angle and a length in the plane. The polar coordinates \(r\) and the angle \(\theta\) can be used to express the Cartesian coordinates:

\[
x = r \sin \theta \\
y = r \cos \theta
\]

**Polarity factor**

*(Radiotherapy)* The current collected from an ionisation chamber, exposed to a constant dose rate changes in magnitude when the polarity of the applied collecting potential is reversed. This is called the polarity effect. The measurement effect can be accounted for by applying a polarity correction factor. For more information on the origin of this effect, please see article Polarity factor.

**Related Article:** Polarity factor

**Polarity effect**

*(Radiotherapy)* The current collected from ionisation chambers exposed to a constant dose rate changes in magnitude when the polarity of the applied collecting potential is reversed. When a new chamber is purchased the effect on its reading of using polarising potentials of opposite polarity should be checked during the commissioning process. For most chambers, this effect may be negligible in photon beams, except for very thin window chambers used for low energy x-rays. This polarity effect is due to the lack of charged particle equilibrium at the collecting electrode. The effect is derived from the relationship of the various primary and secondary interactions occurring within the ionisation chamber. In a photon beam there is a net positive charge near the surface in the build-up region as more electrons are removed from a small mass of material than are deposited and it is reduced to zero when transient electronic equilibrium is achieved. Therefore the magnitude of the polarity effect depends on the thickness of the collecting electrode, the frontal surface area of the collecting electrode, the depth of the collecting electrode beneath the surface of the phantom, the energy of the beam and the field size. The magnitude of the effect depends not only on the maximum energy of the electrons but also on the beam contamination by low energy electrons produced in the collimation system which are stopped in the first millimetre or centimetres. If the chamber is placed at the surface the photon beam initially interacts with the collecting electrode causing electron to be ejected in the forward direction. This results in a region of positive charge being established at the site of the interaction. If a negative bias is applied to the collecting electrode a greater positive charge is collected on the electrode than would be collected from ionisations in the chamber active volume alone. As the depth of the chamber approaches the range of secondary electrons the positive charge due to the ejected secondary electrons is balanced by the negative electrons stopping in the collector. The largest polarity effect is observed at the surface of the phantom since this is the position where the positive charge induced on the collecting electrode by ejection of secondary electrons is not compensated by those electrons scattered from above which stop in the electrode. The effect decreases with increasing depth below the phantom surface as some electrons from the medium above the chamber stop in the collecting electrode. At depths greater than \(d_{eq}\), where a transient charged particle equilibrium exists the polarity ratio reverses but it is less than 1%. For an electron beam the net deposition of charge is positive at the surface from the ejection of \(\delta\) rays and negative at
the depth where the electrons stop in the material. Subsequently the charge deposition from electrons stopping in the medium influences the magnitude of the collected charge which depends on the polarity of the collecting electrode voltage. The average of the opposite polarity readings permit to obtain the true cavity ionisation reading.

The charge is the collected charge using a positive polarity is given by

\[ Q_p = Q - q_1 + q_2 \]

where

- \( q_1 \) is the positive charge created on the collecting electrode due to the ejection of secondary electrons in the forward direction
- \( q_2 \) is the negative charge due to the electrons produced outside the chamber sensitive volume but stopping in the collector

For negative polarity the accumulated charge will be

\[ Q_n = Q + q_1 - q_2 \]

Therefore the true cavity ionisation charge is given by

\[ Q = \frac{|q_1| + |q_2|}{2} \]

The polarity effect is particularly evident in a parallel plate ionisation chamber and can be reduced by the design of the ionisation chamber in particular reducing the electrode thickness. If the polarity effect is significant it is necessary to perform all the measurements with both polarities and to evaluate the average of the two readings.

When an ionisation chamber is calibrated at the standard dosimetry laboratory only one magnitude and sign of polarising potential is normally used. The calibration coefficient therefore refers to that magnitude and sign of the polarising potential. The standard dosimetry laboratories and the manufacturer of the dosimetry systems might refer to the polarity of the dosimeter polarising voltage with different conventions and terminology and therefore, even if these differences do not affect the physics of charge collection in an ionisation chamber, in case the chamber performance with different electrometers is to be compared, special attention has to be paid by the user to the polarity of the polarising voltage.

**Polychromatic beam**

(Radiation Protection) Polychromatic beams consist of several components of radiation. Each component has a specific linear energy transfer (LET). Therefore due to the different absorption of each component, the progressive half value layer (HVL) thickness (namely the thickness of a specific material which reduces the intensity of the radiation entering the material to half) varies. The homogeneity factor (HF) is the ratio between the first and the second HVL and describes the polychromatic nature of the beam. In the case of polychromatic beams, the HF is less than one because the beam is getting harder after each HVL.

**Polycrystalline silicon (Si)**

(Diagnostic Radiology) Silicon (Si) is a metalloid chemical element of atomic number 14 and is a semiconductor. It is used in the construction of integrated circuits which have enabled the evolution of micro-computing and consequently digital diagnostic imaging. Within diagnostic radiology silicon is used in flat panel digital x-ray detectors and flat panel liquid crystal displays, where it forms the principle material in thin film transistors used in active matrix pixel arrays. Silicon generally forms a fourfold, tetrahedrally bonded atomic structure; in crystalline silicon this structure is continued over a long range, whereas, amorphous silicon has no long range order and may sometimes contain dangling bonds as opposed to a pure tetrahedral form. Polycrystalline silicon has a mid range order and consists of arrays of ordered silicon crystals.

Silicon is produce from raw quartzite (silica sand) that purified to create ‘electronic grade’ hydrogenated amorphous silicon. Electronic grade silicon contains less than 0.01 ppm of impurities. Polycrystalline Si is obtained from amorphous silicon by excimer laser annealing (ELA), which allows films of large areas to be manufactured relatively cheaply.

Although amorphous silicon is the most widely used form of silicon, the use of polycrystalline silicon in large area flat panel detectors is a topic of active research due to its superior electrical properties and recent manufacturing advances. Traditionally, amorphous silicon has been used in digital diagnostic imaging as it is easily deposited in thin films over large areas using PECVD (plasma-enhanced chemical vapour deposition). However, although the use of amorphous silicon has been favoured due to its ease of manufacture the electrical properties of crystalline Si are superior. This is due to a higher mobility of charge carriers with the application of an external electrical field. Polycrystalline silicon carrier mobilities are of the order of 100 cm² V⁻¹ s⁻¹, compared to amorphous silicon which has mobilities of approximately 1 cm² V⁻¹ s⁻¹.

As digital imaging resolution is increasing, prototype flat panel imaging systems have been developed which incorporate in-pixel amplifiers constructed from polycrystalline silicon, which increase the signal to noise ratio of the final image. In future developments, by using polycrystalline silicon, it may also be possible for line drivers and readout amplifiers to be integrated on the detector array to reduce external electronics and bring down manufacture costs.

**Related Articles:** Semiconductor detectors, Silicon diode detector, Selenium detector, Silicon, Amorphous silicon


**Polyester in film base**

(Diagnostic Radiology) See Film base

**Polymer gel dosimetry**

(Radiotherapy) Polymer gel dosimetry is based on the free radical chain polymerisation of acrylic monomers dispersed in a gelatin matrix induced by a radiation exposure. The concentration of the cross-linked polymers in a region is proportional to the absorbed dose. The spatially localised polymerisation can then be imaged by MRI or optical scanning methods. When imaged by MRI the relaxation rate of the polymerised region is linearly proportional to absorbed dose in a range relevant for radiotherapy measurements. The change in the optical opacity of the irradiated polymer gel has also led to the development of an optical scan technique for measurements. The optical techniques use a methodology similar to CT scanning but with an optical laser beam instead of a fan x-ray beam. Polymer gel dosimeters present many advantages over gel
dosimeters mainly since there is drastically less diffusion of polymers within the gel matrix so that the radiation-induced changes maintain their spatial integrity. However there are still some disadvantages such as cost of dosimeters, duration of the polymerisation reaction after irradiation and toxicity.

Related Article: Gel dosimetry

Polymer gels
(General) Polymer gels are gels that contain spatially fixed monomers, which polymerise due to radiation exposure. The degree of polymerisation depends on the quantity of free radicals generated by irradiation and therefore the absorbed dose. In this way polymer gels can be used as 3D dosimeters, which are tissue equivalent due to their high water content. Examples of polymer gels include the so-called BANANA, BANG and MAGIC gels.

Polymer gels are found to have greater sensitivity compared to conventional Fricke gel dosimeters. They have sufficient spatial resolution and enable flexible realistic phantom designs. The properties of polymer gels depend critically on conditions, such as temperature and exposure to oxygen and light; therefore, calibration of each gel batch is required.

Polymer gels can be analysed by MRI as water molecules change their binding state and exchange protons with the polymer. An alternative method is optical scanning of the gel's optical density which increases with the degree of polymerisation.

The term polymer gel may sometimes be used to refer to the general class of gels known as hydrogels. Hydrogels consist of a water dispersion medium and water-insoluble polymers with hydrophilic groups that act as the gelling agent. Radiosensitive polymer gels are a specific type of hydrogel.

Medical Applications: Polymer gels are used in medical physics for 3D dosimetry of radiotherapy techniques such as conformal and intensity-modulated radiotherapy, brachytherapy, stereotactic radiosurgery and charged particle radiotherapy.

Related Articles: Fricke dosimeter, Fricke based gel, Gel dosimetry, Gel, Polymer gel dosimetry, Tissue-equivalent material

Porous medium
(General) Porous media consist of a solid matrix containing a network of pores filled with a fluid (i.e. liquid or gas). They can be characterised in terms of their porosity, permeability and the individual properties of the constituent solid and fluid. They generally have a low density, high porosity and a large internal surface area. Examples of naturally occurring porous media include sponge, certain rocks and soils and most biological tissues. Synthetic examples include cements, foams and ceramics. Xerogel is a type of porous medium, which is formed from gels by drying them with an unsolved polymer in the gel matrix.

Medical Applications: Porous materials are used in medicine for their absorptive properties to prevent contamination, for example for wound dressings. In medical physics, MRI can be used to study biological porous media in order to assess flow characteristics.

Related Articles: Ceramics, Gel

Port film
(Radiotherapy) A port film (or portal film) is an x-ray film image taken for verification of external beam radiotherapy. Two types of film exist: those which require 1–3 Gy to give a good exposure and those which require around 50 Gy. The low dose films are used for a pre-treatment set-up check and the high dose films are used to image for the duration of a beam. They are often loosely referred to as check and verification films, respectively. Advantages include high intrinsic spatial resolution. Disadvantages include the need for a film processing unit.

Application: Port film is usually compared with a reference, or gold standard, image showing the planned treatment set-up. This may be a digitally reconstructed radiograph (DRR), a simulation image, or a previous treatment time image. Analysis of treatment accuracy is usually carried out by comparing the positioning of bony anatomy relative to the treatment field edge in the two images.

Alternative Technology: Port film is being replaced by Gafchromic film in many centres, as this obviates the need for the film processor.

EPIDs are also taking the place of portal film for many applications.

Abbreviations: DRR = Digitally reconstructed radiograph and EPID = Electronic portal imaging device.

Related Articles: Gafchromic film, Digitally reconstructed, Electronic portal imaging

Hyperlink: http://www.gafchromic.com/ Gafchromic film

Portal exit dosimetry
(Radiotherapy) Portal exit dosimetry involves evaluation of some aspect of the dose distribution in the patient which is based on measurement of the radiation passing through the patient using a remote imaging detector. The imaging detector used is often an EPID or film. This process involves quantitative evaluation of the pixel intensities in the image (whereas the more common set-up measurement with these imaging systems often involves analysis of the shapes of features).

Calibration: Portal exit dosimetry requires a calibration that relates pixel intensity to dose. This is complex for two main reasons:

1. The detector will generally not be water equivalent and hence have a different dose response as compared to tissue
2. The goal is often to use the measurement in the detector to infer the dose at a point, or a volume within the patient

Hence, portal exit dosimetry often involves complex models that may be empirical or calculation based (e.g. using Monte Carlo methods). Such models often involve corrections for radiation field size, patient/phantom thickness, source to detector distance and the characteristics of the detector (including its linearity with dose).

Plane Dosimetry: One application is plane dosimetry. The plane chosen may be a particular plane through the patient (often the isocentre, or middle of the tissue thickness), the entrance or exit plane, or the plane of the detector. The measurement is often compared with a TPS prediction. In the case of dosimetry in the plane of the detector, the detector is often modelled as an addition slab of material in addition to the patient data.

Volumetric Transit Dosimetry: This involves taking the 2D distribution of intensities in the image and back-projecting through a representation of the patient’s anatomy (often the planning CT scan) to determine a 3D dose distribution. The final process is a mixture of measurement and planning calculation.

Figure P.61 presents a schematic diagram of volumetric transit dosimetry.

Abbreviations: EPID = Electronic portal imaging device, TPS = Treatment planning system and CT = Computed tomography.

Related Articles: Port film, Electronic portal imaging, Dosimetry

**Portal film digitisation**

*Radiography* Portal film is used in many aspects of radiotherapy verification and quality assurance. The advent of software that allows automatic analysis of film data has greatly increased its versatility and ease of use. Such software requires the data in the film to be digitised as an image that can be analysed pixel by pixel.

**Film Scanners:** Specialist medical quality film scanners are available. The most common are probably the systems produced by Vidar (Figure P62). These can scan either a single or stack of films and feed them to an application via a TWAIN interface. With the development of high quality PC document scanners, more mainstream scanners have been used for portal film digitisation with promising results.

**Abbreviation:** TWAIN = Standard for image acquisition devices.

**Related Article:** Port film


**Hyperlink:** Vidar website, [http://www.vidar.com/](http://www.vidar.com/)

**Portal image**

*Radiography* Portal imaging involves acquiring images using a radiotherapy treatment beam. The images are often used to verify treatment accuracy by measuring the positions of anatomical structures relative to the radiation field edge. In many cases the structures used are bony landmarks. More recent attempts have been made to use portal imaging to obtain dosimetric information.

**Comparison with Diagnostic Radiology:** Radiotherapy treatment beams use typical x-ray energies of 4–20 MV. This involves acquiring images using a radiotherapy treatment beam and using these images to verify treatment accuracy by measuring the positions of anatomical structures in the treatment beam. The positions are most commonly measured relative to the edge of the radiation field.

The accuracy of set-up is evaluated in comparison with a reference image, which is usually a digitally reconstructed radiograph (DRR), a simulator image, or a previously acquired portal image.

Two approaches to portal radiography are generally followed:

1. A low dose image to check the set-up accuracy at the start of treatment,
2. An image over a longer time, with a larger dose, during the treatment to verify that the treatment has been delivered accurately.

Abbreviation: DRR = Digitally reconstructed radiograph.

Related Articles: Portal imaging, Port film, Electronic portal imaging

**Position-sensing photomultiplier tubes**

*(Nuclear Medicine)* This is a special photomultiplier (PM) tube design that allows the user to determine the interaction point on the photomultiplier tube surface. These PM tubes are mainly used for nuclear medicine imaging purposes.

Examples of position-sensing photomultiplier tubes are a micro-channel plate multiplier where the electron cloud created in the gamma interaction is confined in a single small channel, and a hybrid photomultiplier tube where the electrons from the photocathode are accelerated onto a corresponding position on a silicon detector.

**Positive contrast media**

*(Magnetic Resonance)* This term refers to contrast agents that make particular tissues more conspicuous by increasing the signal from them.

In MRI, image contrast results from the interplay between the NMR properties of hydrogen nuclei (protons) in tissue and pulse sequence parameters. Positive contrast is usually achieved by using a $T_1$-shortening agent, usually a paramagnetic agent such as a gadolinium chelate, together with a $T_1$-weighted pulse sequence. The $T_1$ of protons in regions receiving a high concentration of the agent are shortened, resulting in faster recovery of longitudinal magnetisation and hence increased signal.

Most contrast agents are administered intravenously, but there are also orally administered positive contrast agents to increase the conspicuity of the bowel. These agents are sometimes based on gadolinium chelates, but also include compounds such as ferric ammonium citrate and oil emulsions (one study demonstrated the utility of ice cream for this application).

Related Articles: Contrast agent, Gadolinium chelate, Paramagnetic contrast agents, Negative contrast media

**Positive-ion cyclotron**

*(Nuclear Medicine)* In a positive-ion cyclotron, positively charged particles are accelerated in circular paths in a magnetic field to very high energies. Particles to accelerate are for example protons or alpha-particles. When the particles have achieved a very high energy they are extracted from the cyclotron using a negatively charged electrode and directed against a target for production of radionuclides.

**Positron decay**

*(General)* Positron decay is a radioactive process in which a proton in the nucleus changes into a neutron and a positively charged electron (positron or beta plus particle, $\beta^+$) is emitted, together with a neutrino, $\nu$: $p^+ \rightarrow n + \beta^+ + \nu$. Gamma rays may also be emitted.
Positron emission tomography (PET) imaging is the abbreviation for positron emission tomography imaging. The radionuclides used in PET are +-emitters. The system is optimised for detecting the annihilation photons and the localisation process takes advantage of the fact that the two annihilation photons are emitted at an angle of 180° to each other. The localisation process is called annihilation coincidence detection (ADC) and with this technique the use of high absorbing collimators is avoided, thus most PET systems have high count rates relative to SPECT and conventional scintigraphy (Figure P.66).

PET Clinical Applications: Conventional x-ray and CT-scanners produce morphological images that give information about the inner structure of the human body. PET on the other hand produces images that describe functional processes inside the body.

There are many clinical uses of PET: oncology (e.g. localisation of tumours and metastases using F18 – FDG as a tracer), mapping human heart function and clinical diagnosis of brain diffusion diseases like Alzheimer’s. It is also used, together with functional magnetic resonance imaging, to map human brain activation.

Today, PET is more often being applied in a combination with a CT or MR-scanner to produce both morphological and functional images. These systems are combined into one camera so they share the same system of reference (even though it is possible to use markers, etc.) and the matching between the two images can be a source of error, for example when using the matched images to select a target volume to irradiate in radiotherapy. A combination between PET and other scanners will minimise the matching source of error and produce an image with both physical and functional information.

Dedicated PET Systems: A number of dedicated PET scanners have been developed and most new scanners have discrete detectors placed in a ring. These scanners have the advantage that they are able to collect all projections simultaneously, thus decreasing the acquisition time and probability for patient movement.

One of the most common scintillation detectors used in PET is BGO which is well suited for this purpose because of its high density and Z-value allowing it to effectively stop high annihilation photons. These detectors are arranged in rings along the trans-axis direction and a PET scanner can have several rings. These scanners can run in either 2D or 3D mode. In a 2D mode the rings are separated by a lead shielding (septa) that prevents photons that are not parallel to the plane of the detector ring to pass. A number of modifications in the 2D mode can be made to increase the sensitivity of the scanner (read more in PET data acquisition article). In 3D mode there are no septa between the rings, this increases the scanner sensitivity but the number of false and scatter coincidences are increased.

Related Article: Data acquisition PET


Post-processing (Diagnostic Radiology) After the reconstruction and corrections for the imaging conditions medical images are often post-processed for further analysis (Figure P.67). Post-processing may consist of several operations such as image enhancement, for example by filtering, segmentation of the image into various regions and co-registration of various images. Also 3D reconstruction of different structures by using surface or volume rendering methods is increasingly used in image analysis and planning of operation or therapy.

In CT imaging many corrections are often made already in connection with the image reconstruction. These include corrections for back-projection artefacts, aliasing, partial volume effect.
and beam hardening. For instance, the beam hardening error can be quite sufficiently compensated for by remapping the projection samples based on known water attenuation characteristics, because more than 80% of the human tissue is water. Some movement artefacts can be corrected for after the reconstruction. Some body movements can be detected and corrected manually, breathing movement can be compensated by holding the breath, but movements affected by the heart can only be corrected by using gated image acquisition and adequate post-processing of the images.

Because the human visual system can only process few grey levels at a time and the monitors are able to show a restricted number of grey levels, the CT image must be windowed. While the whole dynamic range of the image is 2700 HU (Hounsfield units), only for instance 100 HU are displayed at a time, depending on the imaging application.

Image matching of several imaging modalities is mostly based on standard anatomical landmarks. Image matching plays an important role for instance in radiation therapy planning. In this application MRI can give more information about the cancer tissue, while CT data is necessary for calculating the distribution of radiation in the tissues. Combination of several imaging devices, such as PET-CT, helps in co-registration of two modalities.

**Related Articles:** Image segmentation, Gated acquisition


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**Post-processing**

*(Nuclear Medicine)* Post processing refers to the processing of image data following an acquisition. Common post-processing procedures include smoothing, realignment and a number of corrections, for example background and uniformity correction.

**Post-processing** *(Ultrasound)* In diagnostic ultrasound post processing controls are used to alter the grey or colour levels in B-mode images, colour flow images and the spectral Doppler sonogram. By altering the grey level for different echo levels, the appearance of the image is altered *(Figure P.68)* and the difference between specific echo levels can be emphasised. Colour maps can be used for B-mode or spectral Doppler displays. In colour flow imaging, post processing ascribes different colour maps to the CFI map of velocity vectors.

The curves (lower right in each greyscale image) on Figure P.68 show the greyscale level *(y-axis)* against the echo levels received *(x-axis)*. Colour maps are also available on some systems although they are not widely used in clinical practice.

**Posterior** *(General)* Posterior: After, behind, following, towards the rear (e.g. the shoulder blades (scapula) are located on the posterior side of the body).

See also article on *Anatomical relationships*

**Posteroanterior (PA) projection** *(General)* There is a convention where the radiographic technique projection is identified by the direction of the x-ray beam. In the posterior–anterior projection the x-ray tube produces an x-ray beam which passes through the front to the back of the patient to produce an image.

See also *Technique projection*

**Potential difference** *(General)*

**Definition:** In general, potential difference is defined as the difference in potential between two points in a scalar field. In
Potential energy

**Definition and Units:** Potential energy is the energy stored within a particular system, which results from the system's configuration, for example, the position of a charge within an electric field, or a mass in a gravitational field. Potential energy is equal to the work done in moving an object from a reference point to a given position. It is commonly represented by the symbols \( PE, U \) and \( V \). Like all other forms of energy, the SI unit of potential energy is the Joule (J). The force acting on an object \( (F) \) at a point \( (x) \) is related to the potential energy \( (V) \) by

\[
F(x) = -\frac{dV}{dx}
\]

**Examples:** Examples of potential energy include elastic, as in a stretched spring, magnetic, gravitational and electrical potential energy.

**Related Article:** Potential difference

**Power amplifier**

*General* A power amplifier is intended for delivery of power to a load connected to the output stage of a system. In medical imaging systems, power amplifiers are often used to supply actuators for movable parts of the equipment.

**Power breaker**

*General* See Circuit breaker

**Power Doppler**

*Ultrasound* Power Doppler is a generic name for a flow imaging technique where the power, or energy, of the Doppler is displayed, rather than the actual Doppler shift as a measure of velocity. The technique can be used with other trade names such as colour power angio, or energy Doppler. The advantage of power Doppler is that it is more sensitive to slow and small flows than colour Doppler, that is, when velocity is estimated. The same data is used but as the velocity information is disregarded, the gain can be increased by as much as 10dB before the displayed signal is lost in noise. In colour Doppler, the first lag of the autocorrelation is used to calculate the velocity information, but in power Doppler, the zeroth lag is used. Essentially this corresponds to squaring the Doppler time signal, that is \( F + Q^2 \), where \( I \) represents the in-phase signal, and \( Q \) the quadrature component, as obtained from a quadrature demodulation. Another advantage is that power Doppler is virtually angle independent as the power in the Doppler signal is independent of the Doppler angle (as opposed to the Doppler shift). In practice however, flows perpendicular to the sound propagation direction are not displayed properly, due to influence of the wall filter that suppresses low velocities.

That the power of the signal is displayed means, however, that there is a depth dependence, or rather, a dependence of attenuation. The displayed power is also dependent on such factors as gain, transmitted power, pulse repetition frequency/scale and aperture size. This fact has prevented power Doppler of becoming a quantitative tool, and is instead used qualitatively as a means to determine if there is there blood flow present or not. This often gives sufficient information, and has been used to verify flow in kidney transplants for instance. Care should be taken however, since power Doppler is a measure of movement and can give flow images in pulsatile arteries and veins even when there is no net flow through the vessel.

When the technique was introduced in the beginning of the 1990s, approaches were suggested to normalise for the aforementioned dependencies. Since then these have successfully been employed to evaluate flow in fetal lungs (normal, intrauterine growth restricted, and in cases with congenital diaphragmatic hernia), and the brain of normally grown and growth restricted foetuses. Some vendors also offer the total power in a volume of a 3D-data-set as a measure of the amount of flow, as a tool in off-line analysis of recorded data (Figure P.69).

**Related Article:** Angiography

**Power gain**

*General* See Power amplifier

**Power injector**

*Diagnostic Radiology* Power injector is a device used in x-ray angiography to inject contrast media (with specific debit) into the blood vessels and the heart. Normally the contrast media is administered at the required anatomical place through a catheter, which is connected to the power injector. The x-ray generator (as well as the ECG monitor) is connected to the power injector, allowing imaging of specific phases of blood circulation.

**Related Article:** Angiography

**Power spectrum**

*General* The power spectrum gives a plot of the portion of the power (energy per unit time) of a signal against frequency. Most commonly, the power spectrum is calculated by using the discrete Fourier transform. The power spectrum of periodic signals is represented by peaks at discrete frequencies, quasiperiodic signals by peaks at linear combinations...
of irrationally related frequencies. Stochastic signals have a continuous spectrum. Theoretically, one should have an infinitely long sequence of continuous data to calculate the spectrum. Since that is not possible, limitations in calculation accuracy are present due to restricted amount of data and due to sampling frequency (Figure P.70).


**PPI (partial parallel imaging)** *(Magnetic Resonance)* See Partial parallel imaging (PPI)

**Preamplifier** *(General)* A preamplifier is an electronic amplifier at the input of a measurement and processing channel or system. Most preamplifiers have high gain and high input impedance. The preamplifier circuit may have a separate housing positioned near the signal source to be measured.

**Related Article:** Operational amplifier

**Precession** *(Magnetic Resonance)* Precession is the change in the direction of the rotation axis of a rotating object and, in MRI, precession refers to the precession of nuclear spins in the presence of an external magnetic field. The magnetic moment of a nucleus rotates (spins) around its axis and, if positively charged, its spin generates a magnetic field and a magnetic dipole moment parallel to the rotation axis. The magnetic dipole moment can be expressed by a magnetisation vector \( \mathbf{M} \). When placed in a magnetic field \( \mathbf{B} = B_0 \mathbf{n} \), where \( B_0 \) is the magnetic field strength and \( \mathbf{n} \) is the direction of the field, \( \mathbf{M} \) will experience a torque. The equation of motion for \( \mathbf{M} \) is then given by

\[
\frac{d\mathbf{M}}{dt} = \gamma \mathbf{M} \times \mathbf{B} \tag{P.28}
\]

In a time-independent static magnetic field along the \( z \)-axis, the solution to Equation P.28 is given by the so-called Bloch-equations:

\[
M_x(t) = M_x(0)\cos(\omega_0 t) + M_y(0)\sin(\omega_0 t) \\
M_y(t) = -M_x(0)\sin(\omega_0 t) + M_y(0)\cos(\omega_0 t) \\
M_z(t) = M_z(0)
\]

where the Larmor frequency is given by \( \omega_0 = \gamma B_0 \). Above, \( T_1 \) and \( T_2 \) relaxation are neglected.

**Pre-heating of cathode** *(Diagnostic Radiology)* The cathode of an x-ray tube can work at high temperature (necessary for production of the thermal electrons – the anode current) for not more than 1000 working hours. Due to this reason the cathode is heated to this high temperature for a limited time only (during the x-ray exposure). However to heat the cathode from room temperature to more than 2000 K takes significant time. This means that an exposure will start several seconds after the exposure switch is pressed. In order to keep the heating time short the cathode stays always pre-heated to temperature

**Precision** *(Nuclear Medicine)* The measured spread in observations of a measurable quantity is referred to as the system's precision. Precision is also referred to as reproducibility or repeatability. Precision is intimately related to accuracy, which refers to a system's ability to obtain the actual value. As illustrated in Figure P.72 accuracy of a measurement can be high while the precision is low and vice versa.

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**Figure P.71** When a magnetisation vector \( \mathbf{M} \) is the subject of a time-independent magnetic field along the \( z \)-axis, it will start to rotate in the \( xy \) plane with the Larmor frequency.

**Figure P.72** Illustration of precision and accuracy.
around 1500 K. The pre-heating is made by applying a constant stand-by filament current through the cathode (>1 A). This way the time to heat up the filament from the pre-heating to the requested temperature is much shorter (less than a second). When performing radiography the operating x-ray exposure switch has normally two phases (two-steps button). The pre-heating produces some thermal electrons, hence it is important not to switch the high voltage on before the cathode is heated to its high temperature. However some simple dental x-ray equipment allow for switching of the high voltage while the cathode heats up – see Figure V.19 from the article

\textit{Voltage waveform.}

\textbf{Related Articles:} Filament heating, Voltage waveform

\textbf{Preparation (first trigger)}

(Diagnostic Radiology) Usually the exposure switch in x-ray radiography has two stages. The first stage (preparation or first trigger) supplies the necessary filament current to the cathode and rotates the anode to the desired speed (rpm). The second stage (exposure or second trigger) supplies the high voltage to the x-ray tube and produces the exposure.

\textbf{Presampling MTF}

(Diagnostic Radiology) The modulation transfer function (MTF) describes the process of signal amplitude modulation for the different frequencies by the system transferring this signal – in particular the response of an imaging system to an input signal of varying spatial frequency. This modulation in an imaging system is the combined influence of all its components – that is each component has its own MTF. For example in an x-ray imaging system we have the influence of the focal spot > x-ray detector > monitor, etc. (such system is a cascaded linear imaging system). The summary MTF of such a system is the product of the MTFs of all separate components. However this requires that the system is shift invariant (i.e. the image of the object is independent of its position within the imaging field). This assumption is true for analogue imaging systems; however, it cannot be applied in digital systems. This is due to the fact that the imaging chain in such systems includes another function, the digitising of the image. The digitising process samples the signal and divides it into pixels (sampling). According to the Nyquist theorem this inevitably introduces aliasing (i.e. the digital image of an object depends on its spatial frequency content relative to the pixels.). The sampling frequency influences the MTF and we can not use the summary MTF as an overall characteristic of the imaging system. However the MTF of the imaging system before the digitising (the presampling MTF) is not affected by the undersampling phenomenon. The presampling MTF can be used to compare the performance of imaging systems. Normally the presampling MTF contains high spatial frequencies, which otherwise would be affected by the undersampling. The presampling MTF can be measured either directly before the digitising, or by obtaining and analysing the line spread function of the system (LSF). To do this a narrow metal slit is placed at slight angle to the sampling direction and is imaged. The resulting image is a bright line on dark background, produced by the x-rays passing through the metal plate with slit. Each sampling of the image of the line has small offset, relative to the next sampling, what results in reduced sampling gaps (when the samplings are combined) – that is having the effect of increased sampling frequency. This way the resulting presampling MTF includes higher spatial frequencies.

\textbf{Related Articles:} Modulation transfer function, Aliasing, Analogue to digital converter (ADC)

\textbf{Prescribed dose}

(Radiotherapy) The prescribed dose is the dose specified by the radiation oncology team as being appropriate to achieve the purpose of the treatment, for example tumour eradication or palliation, within the bounds of acceptable complications. Dose prescribing and reporting is described in various recommendations, for example ICRU Reports. Prescribed dose should be related to the ICRU reference point. The dose at the ICRU reference point is called the ICRU reference dose and should always be reported.

\textbf{Abbreviation:} ICRU = International Commission on Radiation Units and Measurements.

\textbf{Related Articles:} Treatment planning system, Critical structures, Fractionation, Planning target volume, Normalisation point


\textbf{Pressure and temperature correction factor}

(Radiotherapy) In an ionisation chamber open to the ambient air the mass of air which is inside the cavity chamber is subject to variation. The Standard Laboratories report the calibration factor of an ionisation chamber for reference conditions for pressure, temperature and humidity. If the ambient conditions at the user are different from the standard conditions, the user has to make a correction for the pressure and temperature differences because the ionisation per unit of mass of air varies with the density of air inside the chamber cavity. The correction factor to convert the cavity air mass to the reference condition is given by

\[ f_{p,t} = \frac{273.2 + T(^\circ C)}{273.2 + T_0} \frac{P_0}{P (kPa)} \]

where

\[ P_0 \text{ and } T_0 \text{ are the reference pressure and temperature at the Standard Laboratory} \]

\[ T \text{ and } P \text{ are the ones at the user} \]

Generally the reference value for the pressure is 101.3 kPa and for the temperature is 20°C. The correction factor \( f_{p,t} \) is calculated from the ideal gas law and it is strictly valid without a variation with temperature of the ionisation chamber volume. Chambers with plastic walls may expand their volume with temperature whereas graphite wall chambers exhibit a very small volume variation with temperature. When measurements in a water phantom are performed the chamber waterproof sleeve should be vented to permit a fast equilibrium between the ambient air and the air in the cavity chamber.

\textbf{Pressure parameters}

(Ultrasound) The instantaneous change in pressure at a single field point exerted by a sound wave can be seen as the amplitude waveform signal from a pressure sensitive device such as a hydrophone. Where there is non-linear propagation, the positive going pressure changes may differ significantly from the negative going changes. These amplitudes may be described independently as the peak positive pressure \( P \) and the peak negative pressure \( P \) (Figure P.73).

\textbf{Related Articles:} Intensity, Hydrophone, Acoustic pressure
**Pretargeting**

(Nuclear Medicine) An approach to minimise the radiation dose to normal tissue in patients undergoing radioisotope therapy. Pretargeting refers to the method of sending chemical compounds that target and bind to specific tumour cells. After some time has passed the chemical compounds will have bound to the tumour cells and all excessive chemical compounds are cleared from the body. It is then possible to inject radiolabelled tracer that targets the chemical compounds attached to the tumours. One example of a two-step pretargeting method is to use streptavidin-conjugated primary antibodies that target the tumour. The antibodies are then allowed to bind to the tumour cells and the excess antibodies are cleared from the body. Thereafter radiolabelled biotin can be injected. The radiolabelled biotin binds to the tumour because of the streptavidin-biotin bond.

**Related Articles:** Radionuclide uptake in tumour cells, Extracorporeal elimination


**Primary barrier**

(Radiotherapy) The design of a radiotherapy treatment depends on adequate shielding if the radiation exposure to those outside of the room is to be kept below the regulatory requirements. That part of the room that the primary beam falls on directly is called the primary barrier and will be the most heavily shielded part of the room (e.g. for a 10 MeV linear accelerator this will be of the order of 2.5 m of regular concrete).

Outside of this is a region where lower intensity radiation arising from scattered or leakage radiation falls needs less in the way of shielding and the barrier here is known as the secondary barrier. The thickness of the wall in this region is of the order of 1–1.5 m depending on beam energy (Figure P.74).

**Related Article:** Maze

**Primary collimator**

(Radiotherapy) X-rays are produced in all directions whenever the electron beam impacts on the target in the head of a linac. To provide a useable beam some form of collimation is required to attenuate those x-rays passing through the material. Typically the thickness is such that the leakage radiation is approximately 0.1% of the open field value. The primary collimator is typically a conical-shaped piece of tungsten as illustrated in Figure P.75. The primary collimator defines the largest circular field size available. Further modifications to the beam, for example flattening, further beam shaping, etc. are carried out prior to it exiting the linac head.

**Related Articles:** Linear accelerator, Collimation, Collimator, Treatment head

**Primary colour**

(General) See RGB (red, green, blue)

**Primary radiation**

(Radiation Protection) The primary radiation beam is the x-ray beam directed from the x-ray tube towards the patient. Interactions within the patient lead to scatter and attenuation of the primary beam. The scattered x-rays are referred to as secondary radiation.

**Primary standard**

(Radiation Protection) The primary standard for exposure measurement is a free air ionisation chamber. The x-ray beam or gamma radiation entering the ionisation chamber passing through air interacts (photoelectric effect, Compton effect, pair production) with its molecules. As a result of these interactions electrons and ions are produced and electrical current appears between electrodes. The flow of current is proportional to the exposure rate. The total charge is proportional to the overall exposure. If we assume that KERMA in air ($K_a$) does not differ from absorbed dose in air $D_a$ (Gy), the conversion of exposure in air $D_J$ (C/kg) to $D_a$ (Gy) can be expressed as

$$D_a(J/kg) = D_J(C/kg) \times W(J/C)$$

where $W(J/C) \approx 34$ J/C, the energy to form one ion pair in air.
Then the primary standard can be used for absorbed dose (or dose rate) measurement too.

The accuracy of measurement depends on the electric field distribution (field lines should be normal to the electrodes), and the pressure and temperature of air in the chamber. If the measurement is not performed at STP \((T_0 = 273.15 \, \text{K}, \, P_0 = 1.013 \times 10^5 \, \text{Pa})\) the corrections must be applied.

**Abbreviations:** STP = Standard temperature and pressure and KERMA = Kinetic energy released per mass.

**Related Articles:** Compton effect, Dose, Ionisation chamber, Pair production, Photoelectric effect

**Probability of cell survival**

*(Radiotherapy)* The probability that a cell will survive a dose of radiation generally depends on its radiosensitivity and the magnitude of the dose it receives. This may be obtained from cell survival curves where the probability of survival is given by the surviving fraction as a function of absorbed dose. The linear quadratic model describes this function.

**Related Articles:** Cell survival, Cell survival curve, Linear quadratic (LQ) model, Radiosensitivity, Surviving fraction

**Probability of complications**

*(Radiotherapy)* Radiation treatment inevitably affects normal tissue and so may cause radiation-induced adverse effects. It is generally the case that the total dose that can be tolerated depends on the volume of tissue irradiated – the dose–volume effect. However, the tissue architecture is also thought to be important in determining the tolerance dose for partial organ irradiation (see articles on Parallel organs and Serial organs).

Several radiobiological models have been proposed that relate the probability of normal tissue complications to the dose distribution. For further information see the article on Normal tissue complication probability.

**Related Articles:** Adverse effects, Normal tissue complication probability, Parallel organs, Serial organs, Tolerance

**Production of radiopharmaceuticals**

*(Nuclear Medicine)* The preparation process whereby a radioisotope is labelled to a targeting agent.

The radiopharmaceuticals typically used clinically can be divided into four groups, where each group represents a different labelling technique:

1. Solutions ready for dispensation
2. Prepared kits that only require an addition of for instance \(^{99m}\text{Tc}\)
3. Using prepared kits and an extra effort, for example heating
4. Advanced preparations that involve biological material

Solution ready for dispensation refers to solutions produced outside the department, for example purchased from an external manufacturer or prepared by a dispenser at a pharmacy prior to delivery. There are two basic labelling techniques: ion exchange and introduction of a foreign radioisotope into a molecule. The former technique involves an exchange from a naturally occurring ion to a radioactive ion of the same element. Radiopharmaceuticals prepared this way will have identical bio-kinetic properties as the original radiopharmaceutical. The latter method is used for most \(^{99m}\text{Tc}\) labelling processes. The downside to this is that, due to the introduction of \(^{99m}\text{Tc}\), the final radiopharmaceutical will have different bio-kinetics compared to the radiopharmaceutical it is intended to imitate.

**Production rate of radioactivity**

*(Nuclear Medicine)* See Activation formula

**Projected range**

*(Nuclear Medicine)* The projected range of a given type of charged particle, with a given energy, in an absorbing medium can be defined as the expectation value of the farthest depth of penetration in the particle’s initial direction. In contrast to projected range, the CSDA (continuous slowing down approximation) range describes the true track pattern from starting energy down to rest (Figure P.76).


**Projectile**

*(Magnetic Resonance)* Ferromagnetic metal objects in the presence of the very strong static magnetic fields used in MRI (i.e. up to 60,000 times the earth magnetic field at 3 T) act as projectiles. At reasonable distances from the magnets the field strength falls off according to the dipole approximation at about \(1/r^3\). Long metallic objects show a considerable torque when placed in the static magnetic field and, after the subsequent rotation, their induced dipole moment is parallel to the applied magnetic field. Thus, they experience an attractive force given by \(F = M \cdot \nabla B\), where \(M\) and \(B\) are, respectively, the magnitude of the induced dipole and the magnetic field. In case of a spherical object it has the smallest induced dipole moment and, therefore, the minimum force per unit mass of ferromagnetic material is exerted. The attractive force varies inversely with the seventh power of distance and this results in a sudden increase in the force exercised on a ferromagnetic object brought near the magnet. The force per unit of mass varies inversely with the fourth power of the distance along the axis of the magnet. Accidents where ferromagnetic objects are attracted to the centre of the magnet may result in injury and death of the patient and, therefore, ferromagnetic objects and devices are prohibited in proximity to the MRI scanner. MR safety is based mainly on the construction of physical barriers, metal detectors and the use of strict administrative rules to prohibit accidental introduction of ferromagnetic objects into the magnet room.

**Prone**

*(General)* There are a series of terms used to describe the position of an individual when undertaking different imaging examination. ‘Prone’ means lying on the front.

See also Patient position
Propagation

Propagation describes the passage of ultrasound through tissue. Propagation parameters including speed of sound and attenuation are fundamental to the design and performance of medical ultrasound devices.

**Related Articles:** Speed of sound, Attenuation

**Propagation speed**

(Ultrasound) See Speed of sound

**Proportional counter**

(Radiation Protection) The proportional counter is a gas detector which can operate as a sealed counter with window or a gas flow windowless tube. The voltage applied to the counter is larger than that in an ionisation chamber (Figure P.77). The primary ion pairs, created during the interaction of ionising radiation within the gas and then accelerated by the electric field, create the secondary ionisation process. The gas multiplication $M$ occurs as a Townsend avalanche.

The total charge $Q$ created by $n_0$ primary ion pairs is

$$Q = n_0 \times e \times M$$

where

- $e = 1.6 \times 10^{-19} C$
- $M$ is the gas multiplication ($=10^5$)

The electric field of about $10^6$ V/m is required which for HV of 2 kV can be achieved using a cylindrical geometry (Figure P.78). The anode is a thin wire placed inside a cylindrical or rectangular tube which is a cathode.

The pulse amplitude is proportional to the number of primary ion pairs, i.e. to the energy of radiation. It depends on the kind of radiation. The proportional counters are used to detect x-rays, gamma rays of low energy, $\alpha$ and $\beta^-$ particles, neutrons (detector filled with BF$_3$).

**Protective earth terminal**

(General) See Grounding

**Protective grounding**

(General) See Grounding

**Figure P.77** Dependence of the current (collected charge) on the high voltage value: (a) region of recombination; (b) ionisation chamber region; (c) proportional counter region; (d) limited proportionality region; (e) Geiger–Müller counter region.

**Figure P.78** Scheme of a proportional counter (A – anode; K – cathode).

**Figure P.79** Box proportional counters, side-windows, single: large type PXAr (Kr, Xe) Be38 × 76, small type PXAr (Kr, Xe) Be19 × 38 and double: large type PX2Ar Be38 × 76, different dimensions of cathode body (38 × 76 mm or 19 × 38 mm). (Photo courtesy of Dr. Tadeusz Kowalski.)

**Figure P.80** Box proportional counters, side-windows, single and double. (Photo courtesy of Dr. Tadeusz Kowalski, AGH University of Science and Technology.)
A protocol is a set of guidelines or rules. In nuclear medicine protocols are used to guarantee qualitative examinations and treatments. For example, a protocol can contain guidelines for examination procedures or for quality control of imaging systems.

**Proton**

**(General)**

<table>
<thead>
<tr>
<th>Symbol</th>
<th>P</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mass</td>
<td>$1.672 \times 10^{-27}$ kg (938.27 MeV/c$^2$)</td>
</tr>
<tr>
<td>Charge</td>
<td>$1.602 \times 10^{-19}$ C</td>
</tr>
<tr>
<td>Nuclear spin</td>
<td>$\frac{1}{2}$</td>
</tr>
<tr>
<td>Gyromagnetic ratio</td>
<td>$2.67 \times 10^8$ rad s$^{-1}$ T$^{-1}$</td>
</tr>
<tr>
<td>Radiation weighting factor</td>
<td>5</td>
</tr>
</tbody>
</table>

Protons are stable subatomic particles that were discovered in 1918 by Ernest Rutherford. They are composed of two up quarks and one down quark which are held together by the strong nuclear force. Alongside neutrons, protons form the constituent parts of the atomic nucleus. In magnetic resonance imaging the hydrogen nucleus is often referred to as a proton.

**Medical Applications:** Proton therapy – Due to their Bragg peak, proton beams can offer therapeutic advantage over photon beams in certain clinical situations (most notably in the treatment of anatomically awkward tumours). Proton radiotherapy requires a large initial investment (for a cyclotron or similar accelerator), but the technique is rapidly gaining ground worldwide.

Magnetic resonance imaging – The magnetic moment of the protons in water molecules in the human body is exploited in Magnetic resonance imaging (MRI).

**Related Articles:** Atom, Magnetic resonance imaging, Proton density, Proton therapy, Charged particle therapy, Bragg peak, Cyclotron, Radiation weighting factor

**Proton density**

**(Magnetic Resonance)** One of the parameters modulating the amplitude of the MR signal is the number of the protons (or spins) within the volume of the sample. Certain tissues have more protons per unit volume than other tissues (e.g. water < fat). Voxels with high proton density or hydrogen concentration appear bright. MR signal amplitude depends on the presence or absence of protons (hydrogen nuclei) and is also sensitive to the environment of the hydrogen.
nuclei. In fact it is not only the absolute number of protons in the tissue that is important but also the number of protons within the volume that are sufficiently mobile to be able to line up their spins with the external magnetic field. Tightly bound hydrogen creates a weak MR signal. This effect explains why the cortical bone appears black on an MR image since it does not emit much MR signal. This is not due to the absence of hydrogen but because the hydrogen nuclei are tightly bound to the molecule. On the other hand, medullary bone is visible only indirectly due to the fat located in the space between the trabeculae and in marrow cavities. Consequently, on pure proton density images (i.e. very short TE, very long TR) fat and liquids appear brightest, soft tissue medium grey, and bone dark.

Proton therapy
(Radiotherapy) In proton therapy, high energy proton beams of energy ~250 MeV are used to treat deep seated tumours. The beams are usually produced using high energy proton accelerators such as synchrotrons. They may be shaped using a multileaf collimator (MLC) or by scanning the beam in a raster pattern using a magnetic scanning system. Like other forms of hadron therapy, the dose deposition of protons in tissue displays a Bragg peak, with the majority of the energy deposited in the last few millimetres of the particle's path. Range and energy modulation of the proton beam allows the generation of a spread out Bragg peak (SOBP), which spreads the depth of maximum dose over a region to match the depth extent of the tumour. The depth dose characteristics of a Bragg peak and SOBP are compared with a 6MV x-ray depth dose curve in Figure P.85.

Abbreviation: MLC = Multileaf collimator.

Related Articles: Charged particle therapy, Hadron therapy, Ion therapy, Neutron therapy, Heavy particle beams

Proximal
(General) Near, closer to the origin (example, the proximal end of the femur joins with the pelvic bone).
See also article on Anatomical relationships

Proximal inversion with a control for off-resonance effects (PICORE)
(Magnetic Resonance) The PICORE arterial spin labelling (ASL) preparation technique is a derivative of the EPISTAR concept. The labelling experiment in PICORE is equal to that of EPISTAR. In the control experiment, however, an off-resonance and non-selective pulse, with a frequency offset identical to that of the labelling experiment, is employed (while a slab-selective distal inversion pulse is applied in EPISTAR). Assuming axial imaging of the brain, with known inflow direction, the advantage of PICORE is a minimal contribution to the vascular signal from draining veins (i.e. from inflowing distal spins), with maintained similar cancellation of magnetisation transfer effects. EPSTAR should be considered in cases where eddy currents from slice-selective gradients are significant.

Related Articles: Perfusion imaging, Arterial spin labelling, FAIR, EPSTAR, QUIPSS – QUIPSS II – Q2TIPS

Pseudoecho
(Magnetic Resonance) In conventional 3D MRI, gradient reversal is used to generate an echo which is subjected to Fourier transformation to yield a projection along the frequency encoding gradient direction. The process is repeated many times with a different phase encoding gradient amplitude applied along the orthogonal direction so that the phase of the signal from magnetisation at a given position along this direction is incremented. The phase evolution across the set of acquired echoes mimics the effect of frequency encoding: If the echoes are lined up next to each other in order of phase encoding gradient amplitude, they form a ‘pseudoecho’ along the phase encoding axis that resembles the echo that forms during frequency encoding, and this ‘pseudoecho’ appears to be composed of signals with different frequencies originating from different distances along the phase encoding axis.

Related Articles: B0 gradients, Frequency encoding, Phase encoding

PSF (point spread function)
(General) See Point spread function (PSF)

PSIF (a time reversed FISP)
(Magnetic Resonance) PSIF, the time reversed FISP sequence, belongs to the family of SSFP sequences, a gradient echo sequence combing short repetitions times and low flip angles. However, PSIF is designed to employ the SSFP-echo instead and, maybe, it is not correct to call it a GRE sequence, since it actually behaves more like a spin echo sequence.

The diagram in Figure P.86 shows the timings of a PSIF sequence. By carefully examining the scheme it can be seen that it is a time reversed FISP. Note that in the PSIF sequence, the TE is longer than the TR.

FIGURE P.85 Illustration of the Bragg peak (native proton beam) and spreadout Bragg peak (modified proton beam) for a 250 MeV proton beam in comparison with that for a 6 MV photon beam.

FIGURE P.86 Schematic illustration of a SSFP-echo sequence, PSIF.
The SSFP-echo sequence carries a larger portion of $T_2$-weighted signal, and a combination of short TRs and medium flip angles provides images similar to strongly $T_2$-weighted SEs. In SSFP sequences, which consist of several rapidly repeated RF pulses, the transverse magnetisation from moving spins is spoiled, leading to very low signal from moving spins and signal behaviour similar to the spoiled GRE sequence. If no motion compensation is available, ghosting and signal loss can occur.

The SSFP-echo sequence is strongly $T_2^*$-weighted and has the potential for fast imaging of lesions with long $T_2$ compared with the surrounding tissue. It is presently not widely used clinically.

**Acronyms for the SSFP-Echo Sequences:** Balanced in one direction: CE-GRASS (General Electric), CE-FFE-T2 (Philips) and PSIF (Siemens)

**Related Articles:** Fast imaging with steady state precession (FISP), Spin echo (SE), SSFP, Steady state free precession, $T_2$-weighted


**Public exposure**

*(Radiation Protection)* Public exposures are incurred by members of the public from radiation sources, excluding any occupational or medical exposures and the normal local background radiation but including exposures from authorised sources and practices and from intervention situations.

See also allowed values in **Dose limits**.

**Related Article:** Dose limits

**Further Reading:** IAEA. 1996, International basic safety standards for protection against radiation and for the safety of radiation sources, Safety Series No. 155, International Atomic Energy Agency, Vienna, Austria.

**Pulsatile**

*(Ultrasound)* See **Pulsatility index (PI)**

**Pulsatility index**

*(Ultrasound)* The pulsatility index is a simple, commonly used measure of arterial waveform shape. The index reflects changes in the presence of a proximal stenosis and peripheral resistance (1). The pulsatility index (PI) is non-dimensional and independent of the ultrasound beam/blood flow direction angle:

$$PI = \frac{V_{\text{max}} - V_{\text{min}}}{V_{\text{mean}}}$$

The PI applied to two different shaped waveforms is shown in Figure P.87. The numerator is the total peak-to-peak excursion of the waveform. The mean velocity ($V_{\text{mean}}$) is the time averaged mean of the maximum instantaneous velocity envelope over a number of complete cardiac cycles.

**Abbreviations:** EDV = End diastolic velocity, $V_{\text{min}}$ = Minimum velocity, $V_{\text{max}}$ = Maximum velocity and PSV = Peak systolic velocity.

**Related Articles:** Resistance index, Peak systolic velocity, End diastolic velocity


**Pulse**

*(Magnetic Resonance)* To create MR images of an object a radiofrequency (RF) pulse must be applied. This pulse must have a frequency equal to the Larmor frequency of the spin system. The magnetic field $B_1$ generated by the pulse is rotating in the transverse ($xy$) plane and application of the pulse causes the longitudinal magnetisation to tilt away from the main magnetic field (Figure P.88). The deviation, or flip angle $\theta$ of the longitudinal magnetisation from the main magnetic field depends on the strength and duration of the RF pulse. Directly after the RF pulse is turned off, the longitudinal magnetisation begins to realign with the main magnetic field via relaxation processes.

**Abbreviations:** $M$ is tilted away from the main magnetic field $B_0$ by a flip angle $\theta$.
The duration and amplitude of the RF pulse determines the flip angle. Flip angles between 0° and 90° are often used in gradient echo pulse sequences and a flip angle of 90°, followed by a 180° pulse (spin inversion), is used in a spin echo pulse sequence.

The shape (the time-dependant distribution of amplitudes) of the RF pulse determines the homogeneity of the magnetisation of the spins and the spatial extension of the excitation when a gradient is applied. For example, a sinc (mathematical function defined by $\sin(x)/x$) pulse will ideally excite a slice with a rectangular profile.

**Related Articles:** Flip angle, Larmor frequency, Radiofrequency, Relaxation, Spin echo (SE)

**Pulse average intensity** ($I_{PA}$) *(Ultrasound)* For most diagnostic applications ultrasound is transmitted into the body in a series of pulses whose length and amplitude envelope will depend on the application.

The pulse average intensity is the average intensity of ultrasound over the duration of the pulse (Figure P.89). It is derived from the pulse-intensity integral and the pulse duration. The pulse-intensity integral is the time integral of the instantaneous intensity in the pulse:

$$I_{PA} = \frac{\text{pulse-intensity integral}}{\text{pulse duration}}$$

Or expressed mathematically,

$$I(\text{PA}) = \frac{\int_0^T i(t) \, dt}{T}$$

**Related Articles:** Intensity, Pulse duration, Duty cycle

**Pulse duration** *(Ultrasound)* The pulse duration is the length of time over which the pulse is transmitted. The pulse duration has important consequences for the resolution of the image, the performance of B-mode and Doppler imaging and the time averaged intensity of the ultrasound in tissue.

Pulse duration can be determined from the pulse intensity integral where it is defined as $1.25(t_2 - t_1)$. $t_1$ is the time at which the pulse intensity integral has reached 10% of its final value and $t_2$ is the time at which it reaches 90% of the final value (Figure P.90).

**Pulse echo** *(Ultrasound)* Name of the underlying principle used to generate ultrasound images. This is also the same principle for instance bats and dolphins use to navigate and locate preys. A short sound pulse (for animals we can describe it as a shriek) is emitted and the time until an echo returns is a measure of the distance, provided that the sound speed is known. Human tissue types actually have a range of velocities but the range is fairly small so that an average sound speed of 1540 m/s can be assumed.

**Pulse generator** *(Ultrasound)* A device that outputs pulses, for instance used to trigger a piezo-electric transducer. To emit an ultrasound pulse, a necessary voltage is often in the range of 100 V. To avoid multiple voltage outputs from the transformer, a step-up converter may be used in the circuitry for single element equipments, where the power consumption is lower.

**Pulse inversion** *(Ultrasound)* Pulse inversion is an imaging modality that capitalises on non-linear effects to produce images with improved contrast and/or resolution. The development of the pulse inversion technique was motivated by the limited spatial resolution and image contrast that was attained with harmonic imaging for contrast imaging. Harmonic imaging suffers as the harmonic can only be filtered out over a limited band, otherwise the fundamental energy leaks into the detection bandwidth. This means that only relatively narrow-band pulses can be used in harmonic imaging, and thus a limited spatial resolution is attained.

The solution to this problem as proposed with pulse inversion is to transmit not one pulse, but two. When the echoes from the first pulse have been received, a second pulse, which is a phase-inverted copy of the first, is transmitted (Figure P.91). The two scattered pulses from a linear scatterer will produce two signals that are mirror variants of each other, and their sum will be zero. If, on the other hand, the scatterer is non-linear, as is the case of a contrast bubble, the signals will not sum to zero, but leave a detectable signal. It can be shown that this corresponds to filtering out the fundamental and odd harmonics from
Non-linear scatterers and in some cases the line of response is misplaced. At high count rates two or more events occur in the opposite event in an opposite detector within a mode coincidences occur when an event is followed by a corresponding event in an opposite detector within a coincidence timing window (CTW). Instead, most coincidences registered in such a mode are random coincidences. The data from the delayed mode is used as an estimation of the random coincidence rate and it can be subtracted from images acquired in the image mode.

An obvious drawback is the fact that the coincidences recorded in the delayed mode are not the same as the ones in image mode. Therefore the subtraction of random coincidences will cause an increase in statistical noise.

**Related Articles:** PET, Coincidence timing window, Line of response, Annihilation coincidence detection


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**Pulse pileup**

*(Nuclear Medicine)* Pulse pileup refers to the problem with high count rate detection. At high count rates imaging systems can be incapable of handling the excessive information and separating individual pulses. Pulse pileup effects occur in both single photon imaging and PET but the way the effects are manifested is different in each case.

**Planar Imaging:** Two simultaneously registered events will produce a signal where the signal amplitude (proportional to energy) is the sum of the two individual signals. Using standard pulse-positioning logic the events will be localised as one event somewhere between the two original events. One solution is to have a discriminator circuit that discards all events registering signal amplitude higher or lower than certain thresholds, that is an energy window. But if two photons are Compton-scattered and then simultaneously registered the sum of energy might fall within the boundaries of the energy window, hence creating a false event.

Modern imaging systems use so-called *pulse-tail extrapolation* circuitry. If two photons are registered with only a small difference in time then the signal induced by the first interaction has not fully decayed before the new interaction is registered, hence the signals overlap. It is possible to use estimation circuitry to extrapolate the first pulse and later subtract this estimated signal from the registered signal, theoretically leaving just the signal contribution from the second event. With this method it is possible to decrease the number of pileup events and at the same time the separated events contribute to the image information.

A second approach is available in modern cameras. If the signal distribution of two simultaneously registered events does not overlap they can be separated and read independently from one another.

**PET:** An increase in the count rate will lead to an increase in random coincidences. These random coincidences will degrade the spatial resolution and activity quantification. One method to deal with this problem is the delayed window method. In normal image acquisition mode coincidences occur when an event is followed by a corresponding event in an opposite detector within a coincidence timing window (CTW). At high count rates two or more events occur in the opposite detectors and in some cases the line of response is misplaced.

In the delayed window method, the coincidence timing window is delayed. When an annihilation photon is detected the photon from the corresponding annihilation will arrive within a few nanoseconds at an opposite detector, and if detected it would be considered a true coincidence. When moving the coincidence timing window (~5 times the width of the window itself, the CTW width being typically 12 ns) no true coincidences can be registered. Instead, most coincidences registered in such a mode are random coincidences. The data from the delayed mode is used as an estimation of the random coincidence rate and it can be subtracted from images acquired in the image mode.

An obvious drawback is the fact that the coincidences recorded in the delayed mode are not the same as the ones in image mode. Therefore the subtraction of random coincidences will cause an increase in statistical noise.

**Related Articles:** PET, Coincidence timing window, Line of response, Annihilation coincidence detection

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**Pulse pileup**

*(Ultrasound)* See *Pulse repetition rate*

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**Pulse repetition frequency (PRF)**

*(Ultrasound)* See *Pulse repetition rate*

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**Pulse repetition period**

*(Ultrasound)* See *Pulse repetition rate*

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**Pulse repetition rate**

*(Ultrasound)* Pulse repetition rate (1/s), or pulse repetition frequency is the number of pulses transmitted per second. It can also be referred to as the inverse of the pulse repetition period (Figure P92). The diagnostic ultrasound imaging pulse repetition rate is limited by the depth used as this affects the transmit–receive time. A faster pulse repetition rate gives a higher update frequency of the measured object, which can be relevant to a measurement of a fast developing or moving object.

**Related Article:** Aliasing

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**Pulse sequence**

*(Magnetic Resonance)* The pulse sequence, that is the timing table for the radio-frequency (RF) pulses, magnetic field gradients and signal sampling periods, is one of the most essential links between MR physics and clinical MRI. The possibility to change pulse sequence codes and thereby to significantly alter image contrast is unique for MR as a diagnostic imaging tool, and during the twenty years of MRI in clinical practice that have elapsed, a wide variety of pulse sequences have been suggested and tested.

In order to fully characterise a pulse sequence, normally a graph consisting of five diagrams is used (Figures P93 through P95). These diagrams describe the time frames for execution of RF pulses, gradient pulses and for signal detection (signal sampling period/ADC interval).

![Pulse sequence diagram](https://www.emerald2.eu)

**Related Articles:**

- Physics in Nuclear Medicine
- Ultrasound
- Magnetic Resonance

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**Further Reading:**

The slice thickness can be adjusted by changing the RF bandwidth (the selected frequency interval) and/or the amplitude of the slice selective gradient. Slice-selective RF pulses typically are switched on for a few milliseconds (ms) and their shape is designed to create, ideally, a rectangular slice profile and homogeneous flip angles over the whole width of the slice. Gaussian and, more frequently, sinc pulses are used for this purpose. For the latter pulse type, the slice profile (with a penalty in pulse duration) is improved by adding more lobes to the sinc function. Normally, the slice shape is also improved by using smaller flip angles. By combination of physical gradients in three orthogonal directions, the slice orientation can be freely selected. For maximum signal, it is of great importance to refocus the slice selective gradient: When the slice selection gradient is turned on, spins at different positions in the slice direction have different phase. Since the signal from each volume element (voxel) is obtained by vector summation, so-called spin dephasing and signal loss will be the consequence. In order to compensate for this signal dephasing, a second lobe of the slice-selective gradient with opposite polarity is added after the RF pulse (refocusing gradient). The duration and amplitude of the refocusing gradient should be designed so that the area under the gradient equals the area covered by the original slice-selective gradient pulse, counted from the centre of the RF pulse.

The phase-encoding gradient is switched on after the slice selection and this gradient is used to create spatial resolution by encoding of spin phases in one of the two (orthogonal) directions in the imaging plane. In conventional MRI, the pulse sequence is repeated with different amplitudes of the phase encoding gradient until a sufficient resolution is obtained in the phase-encoding direction, for example 128–256 times. Since the phase-encoding gradient utilises the spin phases to create spatial encoding, it is not possible to prevent spin dephasing in this direction, and therefore the signal will be reduced when large amplitudes of the phase encoding gradient are used.

The frequency encoding (readout) gradient is used to create spatial resolution in the other of the two orthogonal imaging plane directions. This gradient is used to create a spatial frequency dependence of the signal. When the readout gradient is on, a specific frequency interval in the readout direction is defined (readout bandwidth). Typical bandwidth values are a few hundred Hertz (Hz) per picture element (pixel). From the mathematics behind the imaging process, increased duration of the readout interval causes a decrease in bandwidth and an increase in SNR. Reduction of bandwidth, however, enhances chemical shift effects and susceptibility-related image artefacts, for example from metallic implants.

The readout gradient is only used to obtain frequency encoding and hence the phase differences caused by the gradient create an undesired signal loss. Therefore, a refocusing procedure is applied also for the frequency encoding gradient, although the spin refocusing in this case must be made prior to the actual encoding period. This means that the spins are defocused before the signal-sampling period so that full gradient refocusing is achieved at the centre of the signal-sampling period. The encoding period is accompanied by signal sampling (the ADC interval). After the end of the signal sampling period, the sequence is repeated with a new value of the phase-encoding gradient until a sufficient amount of k-space lines have been sampled. After a completed sampling, a 2D Fourier transform (FT) is used to decrypt or decode the raw signal data.
It should be noted that the procedure described previously only covers the basic strategy of pulse sequence design, and that additional RF as well as gradient pulses are frequently used for, e.g. fat saturation, saturation bands, flow compensation and intended spoiling of magnetisation. More extensive overviews of pulse sequence design can be found in Further Reading.

**Related Articles:** Bandwidth, Flip angle, Frequency encoding, Gradient, RF pulse, Sinc function


**Pulse sequence optimisation**

(*Magnetic Resonance*) The MR pulse sequence determines the characteristics of the reconstructed images, for example, contrast, resolution, signal-to-noise ratio, dimensions, location and orientation in a patient, sensitivity for motion. It is thus important that the operator optimises the pulse sequence, that chooses image parameters (such as repetition time, echo time, inversion time and flip angle) to obtain the desired image characteristics with as little distortions and artefacts as possible. This process may implicitly include changing RF or gradient-pulse design and/or timing, and/or changing the readout method. Sequence optimisation can be performed using high-level software tools provided by the manufacturer enabling changes in the sequence protocol, but it can also be made using specific programming languages (pulse programming tools), often available for research purposes.

**Related Articles:** Pulse sequence, Signal-to-noise ratio (SNR)

**Pulse shaping**

(*Ultrasound*) Pulse shaping describes a predistortion of a transmitted pulse in order for it to appear in a desired way either after having propagated a distance, or to appear similar to a pulse produced with a different source (transducer). For instance pulse shaping can be used to minimise unintentional second harmonic transmission in harmonic imaging systems.

A transmitted ultrasound pulse waveform $p(t)$ is the convolution of the excitation waveform $e(t)$ with the transducer impulse response function $h(t)$. In order to produce a transmitted pulse with the desired shape $w(t)$, the transducer must be excited with a voltage waveform:

$$j(t), \ i.e. w(t) = h(t) * j(t).$$

After Fourier transformation of that relation, $j(t)$ can be found as

$$j(t) = i\text{FT} \left\{ \frac{\text{FT}[w(t)]}{\text{FT}[h(t)]} \right\}$$

where $i\text{FT} \{ \}$ represents inverse Fourier transformation $\text{FT} \{ \}$ Fourier transformation

Alternatively it can be found by minimising the quantity $(h(t) * j(t) - w(t))^2$ summed for all time samples.

**Pulsed cine**

(*Diagnostic Radiology*) Pulsed cine mode of operation uses short x-ray exposures for each cine frame. During this mode the x-ray beam switches rapidly on–off, thus producing sequence of images (most often with rate of 25–30fps). Using pulsed cine mode requires special design of the x-ray generator (most often associated with use of grid controlled x-ray tube). The exposures have to be synchronised with the cine camera shutter. Often such mode can also be synchronised with the ECG of the patient, thus filming specific phases of the cardiac cycle (in cardio angiography). The fact that the x-rays are off between two frames leads to an overall reduced patient dose, compared with continuous mode cineangiography.

**Related Articles:** Grid controlled x-ray tube, Cineradiography

**Pulsed dose rate (PDR)**

(*Radiotherapy, Brachytherapy*)

**Dose Rates in Brachytherapy:** Different dose rates are used in brachytherapy treatment techniques. The International Commission on Radiation Units and Measurements, ICRU, defined these dose rates in its Report No. 38 ‘Dose and Volume Specification for Reporting Intracavitary Therapy in Gynecology’:

1. Low dose rate, LDR
   a. 0.4–2.0 Gy/h
   b. Traditional technique; 0.5 Gy/h, 60 Gy with treatment time 5 days
   c. Large amount of clinical data
   d. (NOTE: Ultra low dose rate 0.01–0.3 Gy/h)
2. Medium dose rate, MDR
   a. 2–12 Gy/h
   b. Seldom used
3. High dose rate, HDR
   a. >12 Gy/h = 0.2 Gy/min
   b. Treatment times approximately 5–20 min (comparable to external beam therapy)
   c. Clinical data available
4. Pulsed dose rate, PDR
   a. Mimics LDR, using many small ‘HDR pulses’ during a longer treatment time

Example: 1 Pulse per hour during 24 h, 0.5 Gy per pulse given in 5 min; total dose 12 Gy/day.

The radiobiological effects in the tissues irradiated depend on the type of applicator used, on the fractionation scheme and on both dose and dose rate distributions. As stated in the ICRU Report 38: ‘the clinical experience accumulated with radium techniques cannot be applied to new irradiation conditions without careful consideration’. This includes consideration of both tumour effects and effects on normal tissues.

**Abbreviation:** ICRU = International Commission on Radiation Units and Measurements.

**Related Articles:** Brachytherapy, Dose rates in brachytherapy, see also articles under radiobiology

**Further Reading:** ICRU. 1985. Dose and volume specification for reporting intracavitary therapy in gynecology. ICRU Report 38, ICRU, Washington, DC.

**Pulsed mode**

(*Diagnostic Radiology*) Pulsed mode of operation is used in some contemporary x-ray fluoroscopic systems. During this mode the x-ray beam switches rapidly on–off, thus producing sequence of images. Each image is recorded in the system memory. Using pulsed mode requires special design of the x-ray generator (most often associated with use of grid controlled x-ray tube). The resulting images are with rate of 25–30fps, or more. The fact that the x-rays are off between two frames leads to an overall reduced patient dose, compared with continuous mode fluoroscopy.
Pulsed mode fluoroscopy has another advantage, related to image quality. Due to the short exposures of each frame the resulting images have less motion artefacts from movement of the anatomical organs – that is these are sharper (with better resolution and less noise).

**Related Article:** Grid controlled x-ray tube

**Further Reading:** Beutel, J., H. Kundel, R. van Metter, eds. 2000. *Handbook of Medical Imaging, Volume 1, Physics and Psychophysics*, SPIE Press, Washington, DC.

**Pulsed mode**

(*Ultrasound*) Pulsed mode describes a mode of operation where the ultrasound is transmitted intermittently, that is in bursts or pulses, by the transducer. Between each pulse, the echoes from targets within the body are collected. Using the time between the transmitted pulse and the received signals, the distance at which the echoes were generated can be determined:

\[ d = \frac{c \times t}{2} \]

where

- \(d\) is the distance of the target from the transducer
- \(c\) is the speed of sound in the tissue
- \(t\) is the round trip time for the burst of ultrasound

Example: A transducer is placed on the surface of the body over tissue in which the speed of sound is 1540 m/s. An echo is received \(1 \times 10^{-4}\) s after the transmitted ultrasound pulse. The depth of the structure that generated this echo is \((1540 \times 10^{-4})/2 = 0.077\) m or 7.7 cm.

All ultrasound imaging and almost all other diagnostic applications of ultrasound rely on pulsed mode for their function.

**Related Articles:** B-mode, Pulsed ultrasound

**Pulsed radiation**

(*Diagnostic Radiology*) See the article on Pulsed cine mode

**Pulsed ultrasound**

(*Ultrasound*) Pulsed ultrasound uses the emission of a time-limited burst of sound, as opposed to continuous wave ultrasound. With pulsed ultrasound, range detection of echoes is possible and pulse echo techniques are the basis of B-mode imaging. Pulsed ultrasound is also used for pulsed Doppler, using slightly longer pulses where the phase shift between successively received echoes is sensed. This forms the basis of colour flow imaging and pulsed wave spectral Doppler in clinical ultrasound scanners.

**Related Articles:** B-mode, Pulsed mode, Pulsed wave Doppler, Colour flow imaging

**Pulsed wave Doppler**

(*Ultrasound*) Pulsed wave Doppler techniques are used for colour flow imaging and spectral Doppler in ultrasound scanners (Figure P.96). By transmitting pulses and measuring phase shifts between successive echoes to reconstruct the Doppler shifts, pulsed Doppler can examine movement at particular depths. The length of the pulse and the interval over which analysis is undertaken dictates the axial resolution, the time between pulse transmitted and received echoes determines the depth at which analysis is made.

The processing of pulsed wave Doppler is different for spectral Doppler and colour flow imaging and reflects the requirements for each. For colour flow imaging, rapid calculation of movement in several sample volumes along each of several lines is undertaken by auto or cross-correlation. For spectral Doppler displays, sample and hold of narrowband pulses phase shifts are used to reconstruct the Doppler shift for audio output and onward processing for the sonogram.

Pulsed Doppler suffers from one limitation which has profound consequences for colour flow imaging and spectral Doppler. The maximum Doppler shift that can be measured unambiguously is half the pulse repetition frequency (PRF), the Nyquist limit. This limit can be almost doubled if flow is unidirectional but the frequencies used for ultrasound, blood velocities in the arterial circulation and depths in the abdomen combine to provide practical constraints. If Doppler frequencies exceed half the PRF then aliasing occurs whereby the displayed flow appears to be in the wrong direction, either on colour flow or spectral Doppler (Figure P.97). This can be useful, for example in identifying areas where blood suddenly accelerates at a narrowing but can lead to difficulties in measuring high
In differential mode the PHA selects pulses by two levels (height window): lower level (baseline) \( V \) and a window size \( \Delta V \) (upper level). Only the pulses with amplitude \( (V, V + \Delta V) \) pass to the recording device. In integral counting all pulses with a height above a certain value \( V \), selected with the lower level discriminator, are counted.

The measuring system is calibrated, i.e. the proportionality between a pulse height \( V \) and energy \( E \) of detected radiation is known, and with use of PHA one can separate counting corresponding to the chosen energy of radiation. If it is for only a single energy, for example a photopeak the PHA is called a single-channel analyser (SCA).

**Abbreviations:** PHA = Pulse-height analyser and SCA = Single-channel analyser.

**Related Articles:** Proportional counter, Scintillation detector, Semiconductor detector, Single-channel analyser

**Further Readings:**

**Pulse-less generator**

(Diagnostic Radiology) The pulse-less x-ray generator (also called constant potential generator), is usually a 12 pulse x-ray generator with high voltage stabiliser (often using high voltage tetrodes or triodes as regulating elements). Figure P.100 shows a typical block diagram of such a generator. It includes a feedback circuit with high voltage dividers, which supply the regulating circuit with signal proportional to the kV across the x-ray tube. This signal is compared with a reference value (the set kV) and, in case of difference between these, changes the regulating voltage of the tetrodes/triodes, which regulates the high voltage over the tube.

The pulse-less x-ray generator was the highest level of development of the classical x-ray High voltage generator. It was used for powerful x-ray equipment (as angiographs) and CT scanners. After the 1990s all these are gradually being replaced by medium frequency x-ray generators.

**Related Article:** High voltage generator

**Pulse-pressure-squared integral**

(Ultrasound) The time integral of the square of the instantaneous acoustic pressure at a particular point in an acoustic field integrated over the acoustic pulse waveform (IEC 61157). Unit: Pa²s

\[
p_{p} = \int_{\text{the pulse}} p^{2} \, dt
\]

**Related Articles:** Beam width, Intensity

PVDF
(Ultrasound) Polyvinylidene fluoride, or PVDF is a highly non-reactive and pure thermoplastic fluoropolymer.
In ultrasound, PVDF is used as a membrane in hydrophones.
For further information see Hydrophone.

PZT
(Ultrasound) PZT, lead zirconate titanate, is the most commonly used transducer material.
For further information see Transducer.
**Q factor**  
*(Diagnostic Radiology)* Sometimes image quality parameters and related dose parameters can be linked in a formula, presenting the quality as a single figure. This new empirical figure is an imaging performance parameter, known as $Q$ factor or $Q$ value. However it is not an absolute figure, as it often depends on additional parameters, such as type of test object, x-ray system parameters, etc. For example, one such $Q$ value, used for the evaluation of some CT scanners, includes a formula incorporating the 50% value of the MTF, the percentage of image noise, the nominal slice thickness and the Dose in water.

**QC (quality control)**  
*(General)* See various articles on the subject under the name Quality control

**QDE (detective quantum efficiency)**  
*(Diagnostic Radiology)* See Detective quantum efficiency (DQE)

**Q-space (used in diffusion-MRI and NMR)**  
*(Magnetic Resonance)* The $q$-space imaging technique utilises extremely high diffusion sensitivities which makes it possible to study the geometrical shape and extension of several chemical compounds, such as polymers and emulsions. The technique has been in use in the field of nuclear magnetic resonance (NMR) for a long time but has not been applicable within the area of MRI due to limitations of the MRI scanner hardware, mainly related to the maximum achievable gradient strength. With increasing system performance it has now become possible to achieve higher diffusion sensitivities also in MRI, however, not as high as in the NMR environment (due to physiological limits like peripheral nerve stimulation).

The technique is based on the same method as for conventional diffusion weighted (DW) MRI, but includes a large number of measuring points with continuously increasing sensitivity where the maximum sensitivity is inversely proportional to the achievable $q$-space resolution. In $q$-space imaging the diffusion sensitisation is described by the wave vector, or the $q$-value, $q = \frac{\gamma}{2\pi}G\delta$ [m$^{-1}$], where $\gamma$ is the gyromagnetic constant, $\delta$ the gradient duration and $G$ the gradient amplitude. Measurements are often carried out using several different diffusion times and this allows for studies of restricted or hindered diffusion.

In MRI the $q$-space imaging method is used to obtain a probability displacement distribution, achievable from the Fourier transform (FT) of the signal decay curve. Thereby the diffusion can be quantified with the mean displacement or full width at half maximum (FWHM) of the displacement distribution, a measure which is dependent upon the diffusion time.

In the figure a FWHM map obtained from $q$-space imaging is seen. The image contrast is similar to an apparent diffusion coefficient (ADC) map, however, depending upon the measuring time, see ADC.

**Related Article:** Apparent diffusion coefficient (ADC)

**Quadrature**  
*(Magnetic Resonance)* See Phase quadrature

**Quadrature artefact**  
*(Magnetic Resonance)* RF quadrature artefacts are caused by a disturbance in the two detector channels of the quadrature detector. DC offset of the output of one of the amplifiers will produce a bright point in the centre of the image. A higher gain of one detector than the other will result in a ghost appearing diagonally opposite from the real object.

These artefacts can be caused by the technical faults within the quadrature detectors and typically must be rectified by the service engineer.

**Abbreviations:** RF = Radiofrequency and DC = Direct current.  
**Related Article:** Ghost artefact

**Quadrature coil**  
*(Magnetic Resonance)* An RF quadrature coil (also called a circularly polarised or CP coil) is a coil that is used to transmit and/or receive a circularly polarised $B_1$ field.

A circularly polarised RF transmit coil requires only half the transmit power required by an equivalent linearly polarised coil. A quadrature coil creates a spinning $B_1$ component that rotates in the same sense as the precession of spins.

The conceptual operation of a quadrature transmit coil is shown in Figure Q.1. The driving signal from the RF generator is split
Quadrature detection

Quadrature detector

into two paths. A phase shifter creates a 90° phase shift between each path. The geometry of the coil is such that each of the driving signals generates mutually orthogonal linearly polarised magnetic fields. As these component fields are both orthogonal and in quadrature, the resultant is a circularly polarised field.

Quadrature coils also operate as RF receivers. The signal to be detected (the spinning transverse component of magnetisation) is circularly polarised. Again, this spinning component can be thought of as two orthogonal, in quadrature linear polarisations. A linearly polarised receive coil would detect only one of these polarisations. A quadrature receive coil is more efficient, detecting both polarisations and giving an SNR improvement of a factor of $\sqrt{2}$ compared to an equivalent linearly polarised coil. The reception process is essentially the reverse of the transmission process shown in Figure Q.1. The geometry of the coil is such that one received signal represents a linear polarisation of the spinning transverse component and the other represents the orthogonal linear polarisation. The two signals detected are 90° out of phase and are shifted relative to one another prior to recombination.

**Quadrature detection**

**(Ultrasound)** In quadrature detection, the received signal is mixed with two different signals, one in-phase component (cosine), and one quadrature component being 90° out of phase with the first (sine). Usually these components are referred to as the I-channel (or component) and the Q-channel (or component). This phase shift can for instance be accomplished with a high reactance capacitor. The reason for employing quadrature detection in Doppler systems is that the reverse and forward flow components can be separated, using additional circuitry shown in Figure Q.2. The 90° phase shift after the low-pass filter (which is there to remove the sum frequency components) is, however, difficult to achieve over the whole range of Doppler frequencies using analogue components. Cascaded RC-circuits may be used but if the signal can be digitised, a digital Hilbert transform can be employed instead. The Hilbert transform performs a 90° phase shift of all frequencies up to the Nyquist limit. For the separation of reverse and forward components to be successful it is very important the phase shift is exactly 90°. If not, a spurious component will show up at the mirrored frequency, i.e. if only a forward Doppler shift of 400 Hz is evident, a weak reverse shift at −400 will also be apparent. Less than 2° of mismatch will keep the spurious component under 36 dB compared to the wanted signal. A similar error can arise from having unmatched amplitudes in the low pass filters after the mixers.

**Related Articles:** CW Doppler, Doppler ultrasound

**Quadrature detector**

**(Magnetic Resonance)** Quadrature detection is a signal processing method for extracting in phase and in quadrature signal components from an input signal, relative to some reference signal. The process is illustrated in Figure Q.3.

In MRI, the ‘reference signal’ originating from the RF oscillator can be written as

$$S_{\text{ref}}(t) = A_c \cos \omega_c t$$  \hspace{1cm} (Q.1)

This signal is driven with a frequency $\omega_c$, corresponding to the Larmor frequency.

The signal component detected by the RF receiver coil from any given voxel is

$$S_{\text{detect}}(t) = A \cos ((\omega_c + \Delta \omega) t + \phi)$$  \hspace{1cm} (Q.2)

**FIGURE Q.1** Quadrature coil in transmission. The coil is shown looking down the z-axis (i.e. along the bore in a conventional MRI).

**FIGURE Q.2** Block diagram of a unit for separation of forward and reverse Doppler components using quadrature detection.

**FIGURE Q.3** Quadrature detection.

Quality assurance

(Magnetic Resonance) Quality assurance is the management and review of the quality control tests that should be performed on a regular basis on MRI equipment. The aim of quality assurance is to ensure that the quality control tests take place, the protocols and results are reviewed regularly and any necessary alterations are made to the procedure.

Legal requirements to fulfil specific criteria in MRI quality assurance will vary from country to country but a best practice policy should be adopted to make sure quality assurance takes place. This will ensure that the equipment is maintained at a clinically useful level.

Related Article: Quality control

Quality assurance

(Radiation Protection; General) Quality assurance (QA) is the system that provides confidence in the safety, reliability and accuracy of a medical device or process. A QA system is based on a set of procedures that allow the monitoring of the medical device or process, such that if a fault occurred it would be quickly noticed and the equipment removed from service/process suspended for review.

A QA system should encompass the life-cycle of a medical device/process, instilling quality principles at specification, tendering, acceptance, commissioning, in-service and de-commissioning stages. A mature QA system should inform the specification and tendering process, just as acceptance and commissioning should inform the in-service QA.

QA systems generally categorise faults relative to their importance, with patient-safety foremost, followed by lower-level faults relating to equipment reliability and/or data-accuracy. Patient-safety is of paramount importance to most equipment vendors (their sales depend on it), and safety issues should have been exhaustively tested by the manufacturer. Nevertheless, safety-tests form part of routine QA procedures; these tests are designed to check patient-critical systems which may not be encountered during routine clinical operation. Some equipment faults may be spotted first by clinical staff, and so fault-logs also form part of the QA system.

Most QA activity is taken up by assessment of accuracy/reliability of equipment/processes, ensuring that data produced by a medical device/process is accurate. The tests undertaken to determine accuracy and reliability are defined under quality control (QC) procedures, which form part of the overall QA system.

Process optimisation is another aspect of QA that aims to reduce errors in clinical procedures. This is of particular importance in areas such as diagnostic imaging, where a variety of clinical and technical factors can invalidate the usefulness of a scan. As well as the economic impact of poorly optimised procedures, there are additional risks associated with ionising radiation that necessitate a ‘right-first-time’ ethos. Thus process review, QC results and fault-logs are all aspects that contribute to a responsive QA system.

Related Article: Quality control

Quality assurance

(Ultrasound) Quality assurance is a term used to describe methods and processes to ensure that products and services are of a sufficient standard for their purpose. It is commonly used in manufacturing where there are internationally agreed terms to cover various aspects of the quality process under ISO 9000 (www.iso.org).
In clinical ultrasound the term quality assurance is commonly used to describe performance testing of ultrasound equipment. This may be done at various stages including:

- Comparative testing and evaluation prior to purchase
- Acceptance testing prior to clinical use
- Continued performance assessment in service

Assessment of system performance can be performed clinically (e.g. is a system adequate for its task and does it remain so) or by technical assessment of performance, usually using ultrasound phantoms (link).

Change in performance can often be gauged by users, for example if controls do not work or work inadequately or if there are obvious changes in the image appearance or new artefacts appears. More subtle changes in performance may be difficult to observe in everyday use. The need for routine quality assurance is controversial. Many protocols were designed for systems with a high proportion of analogue circuitry; the increased use of digital processing and the ability for self-testing means that beamforming and the processing of images is less likely to suffer degradation in performance. The transducer is still at risk from deterioration or damage and some scientists argue that the focus of QA testing should be on transducer performance.

A large range of measurements can be made, for example in B-mode:

- Spatial resolution axial, lateral, elevation
- Penetration, (sensitivity)
- Contrast resolution
- Measurement accuracy (callipers), linear and area
- Image uniformity (includes monitor)
- Dead zone
- Hard copy performance

Doppler parameters include:

- Spectral Doppler
- Velocity direction indication
- Sample volume dimensions
- Directional discrimination
- Penetration depth (sensitivity)
- Velocity estimation accuracy
- Waveform index estimation accuracy
- Volume flow estimation accuracy

Colour flow imaging
- Lowest detectable velocity
- Highest detectable velocity
- Image spatial resolution (and registration accuracy)
- Temporal resolution
- Velocity resolution
- Tissue colour suppression performance

In practice, making consistent measurements from phantoms require considerable expertise, spatial resolution parameters, for example require rigorous attention to detail for the environment, scanner settings and use of the phantom. Doppler parameters are particularly difficult to measure and a simple measurement such as peak velocity is dependent on depth, power, gain, beam steering and position in the array.

A diagrammatic representation of ultrasound transducer and phantom and the resulting image is shown in Figure Q.4.

The insonated region on Figure Q.4 is shown by the box on the side of the phantom beneath the transducer face. Images of the wires show as a series of points permitting measurement of axial and lateral resolution and calliper accuracy.
Quality audit

(Radiation Protection; General) An assessment of the systems and procedures and the adherence to those procedures within a department using radiation. An assessment of systems and procedures would check compliance against legal requirements (e.g. national ionising radiation regulations) as well as accepted codes of practice. Assessing adherence to local protocol is achieved through various quality control metrics, such as auditing patient administered doses, image quality and equipment maintenance.

There must be appropriate mechanisms in place to review the results of quality audits, to feed back those results and apply corrective actions, and to ensure the actions are completed.

Related Articles: Quality control, Quality assurance

Quality control

(Magnetic Resonance) A quality control system is put into place to ensure that the MRI system is maintained at a clinically useful level. System variability can be caused by drift in the electronics of the system, slow drift of the magnetic field, gradient system failures or the introduction of ferromagnetic articles into the scanner room, for example. These factors can lead to a variety of artefacts or changes in image characteristics. The most important of these arguably is a decrease in SNR, caused either by a reduction in signal or an increase in noise. The reduction of SNR will lead to a less clinically useful image which, if not noticed, may affect clinical diagnosis.

A clear schedule of quality control measures is important, with the most important measures repeated more frequently. Typically, a daily test is performed with a uniform test object and a standard pulse sequence to inspect the SNR and image uniformity. This gives a good overview of the performance of the scanner in a short amount of time. The SNR can be monitored over time to look for any trends or sudden problems in the scanner.

Other tests are performed on a less frequent basis to provide a more in-depth and detailed analysis of the different parameters that can be affected by system variability. These include SNR, spatial resolution, uniformity, geometric distortion, slice thickness, slice position, slice separation and ghosting. This set of tests requires a set of appropriate test objects.

Some quality control measures may be specific to the clinical and research interests of the site, for example breast imaging or MR spectroscopy.

Abbreviation: SNR = Signal to noise ratio.

Quality control

(Nuclear Medicine) Quality control refers to the activities performed to acquire evidence to verify quality in specific parts of a business or department.

In nuclear medicine, quality control refers to the systematic control of imaging systems, patient and personnel routines to ensure that the service fulfills the predefined quality conditions. A specific quality control can be a step in the QA process where the entire research and clinical activity are reviewed and tested.

One example of a quality control routinely preformed is the uniformity measurements of a scintillation camera.

Related Articles: Quality assurance, Uniformity

Quality control

(General) Quality control (QC) refers to the processes by which the data-accuracy and safety of a medical device is assessed. Quality control describes any set of tests that should be carried out on an equipment to determine whether it is fit for purpose.

Generally, a minimum set of tests that should be carried out on a piece of equipment is defined by a professional body or a national regulatory authority. These will set-out required tests and the acceptable performance for each test. Acceptable performance may be defined in terms of absolute accuracy, or a tolerance of variation from a previous measurement (e.g. at commissioning).

QC then concerns itself with monitoring of data-accuracy over the in-service use of the medical device, and informs the QA system, which encompasses the broader clinical application of the device, and all potential-related sources of error, over the life-cycle of the device.

Related Article: Quality assurance

Quality control

(Ultrasound) Quality control is part of the larger quality assurance process of a product or service. In manufacturing, quality control could include monitoring and auditing the manufacturing process and product. Quality assurance would include the larger process including a company’s systems to ensure the design and manufacturer of systems lead to a high quality product. In industry, quality is addressed by the ISO 9000 group of standards for quality management systems.

In ultrasound imaging the term quality control might include the testing of scanners to ensure that they achieve the designed standards of performance and safety. The term is rarely used in clinical applications and departments where the phrase ‘quality assurance’ is used when describing ways to ensure the performance of scanners in clinical use. The use of this term is discussed under quality assurance.

Quality factor

(Magnetic Resonance) The equivalent circuit for a simple RF coil is shown in Figure Q.5a. The resistance represents the resistance of the unloaded coil plus loading effects of the patient.

The frequency response Figure Q.5b shows the variation of the current flowing in this coil with applied driving frequency. The coil shows a resonant peak at a frequency \( \omega_0 \). The quality factor or Q-factor is a measure of the sharpness of this peak.

For the series circuit in Figure Q.5b the quality factor is given by

\[
Q = \frac{\omega_0 L}{R}
\]

The quality factor can also be calculated from the measured coil frequency response:

\[
Q = \frac{\omega_0}{\Delta \omega}
\]

where \( \Delta \omega \) is the measured coil bandwidth.

The Q-factor is an important measure of coil performance as SNR improves with the square root of \( Q \).

\[
\text{SNR} \propto \sqrt{Q}
\]

![Figure Q.5](image)

(a) Equivalent circuit for an RF coil. (b) Frequency response of coil. The bandwidth is the frequency difference between the frequencies where the response has fallen to −3 dB of the peak response (i.e. reduced in amplitude by a factor of \( \sqrt{2} \)).
Quality Index
(Radiotherapy) Linear accelerator beams tend to be generically specified by a nominal accelerating potential (e.g. 4, 6MV, etc.). While this aids the user in a simple definition of the treatment beam, it by no means specifies the true nature or energy of the beam. This is important as two machines with the same nominal accelerating potential can produce very different beam properties such as percentage depth doses, output factors, etc. and the user must be aware of this before thinking about beam matching.

While in diagnostic radiology a beam can be specified in terms of half value layers (HVLs) of materials such as copper or aluminium, the spectrum of energies in a linear accelerator treatment beam renders this beam qualifier irrelevant. The beam specifier of choice in radiotherapy is called the quality index.

The quality index is simply a measurement of the tissue phantom ratio (TPR) at two specific depths, 10 and 20 cm, and this is often denoted by TPR1020. The measurement is made with the chamber positioned at the isocentre for both readings, and with a set field size of 10 × 10 cm². An illustration of the setup for each field is given in Figure Q.6.

A measurement of the quality index is essential for current absorbed dose to water codes of practice for calibration of MV treatment beams. Typical values are 0.656 for 6MV and 0.758 for 15MV beams.

Alternatively, the quality index Q can also be determined from depth dose measurements with fixed SSD of 100 cm at depths 10 cm and 20 cm with a formula derived from experimental data by Andreo et al. (1986):

\[ Q = 2.189 - 1.308 \times j + 0.249 \times j^2 \]

with \( j = \frac{M_{\text{ill}}}{M_{\text{cin}}} \).

Related Articles: Calculation of absorbed dose, Beam energy, Tissue phantom ratio (TPR), Beam quality


Quantum detection efficiency (QDE)
(Diagnostic Radiology) See Detective quantum efficiency (DQE)

Quantum efficiency
(Diagnostic Radiology) The quantum efficiency can generally be defined as the percentage number of input quanta that contribute to the output. A quantum efficiency can be defined for each individual step in an imaging process where photon or charged particle interactions take place and often the input energy is converted from one form to another.

For example, in the photocathode of a photomultiplier, the quantum efficiency is

\[ \text{QE} = \frac{\text{number of photoelectrons emitted}}{\text{number of incident light photons}} \]

The QE may be expressed as a single figure for a quantum interface or, since the interactions will often have an energy dependency, can be expressed as a function varying with energy or wavelength. The lower the quantum efficiency, the lower the sensitivity of detection and this leads to a higher noise component. In a chain of conversion steps, a particularly low quantum efficiency is described as a quantum sink.

Quantum mottle
(Diagnostic Radiology) Quantum mottle refers to the pattern of quantum noise, or random variations, in a radiographic image which is due to the statistical fluctuation of x-ray photon absorption and consequent light photon emission by an intensifying screen or digital radiology scintillator screen. The faster the intensifying screen, or the higher the kV, the more light photons are produced and so fewer x-ray photons actually contribute to a final image of a desired optical density or pixel value.

Since the emission and detection of photons are normally distributed, the noise, or randomness, associated with the number of photons is proportional to the square root of the number of photons. So with fewer x-ray photons contributing to the image, the greater is the noise, or mottle, seen in the image.

Quantum noise
(Nuclear Medicine) Quantum noise refers to the fluctuations in detected signal because of the finite number of discrete signal (energy) carriers (e.g. photons, electrons). These fluctuations are manifested as noise in an image. By measuring a known intensity the average number of photons collected can be established but will not lead to knowledge of the actual number of photons collected per element (pixel). The number of photons collected per element will have a mean value in accord with a Poisson distribution (Poisson statistics).

Related Articles: Pixel, Poisson distribution

Quantum number
(Nuclear Medicine) Quantum numbers are values describing the dynamics of a quantum system such as an atom or a particle. Quantum numbers can describe the energy state of an electron in an atomic system or the angular momentum or spin of an electron. The electron orbits in an atomic structure is described by four quantum numbers: principle (n), angular (l), magnetic quantum (m) and spin (s) quantum number. n, l and m describe the size, shape, orientation of the orbit respectively, s describes the spin of the individual electrons.

Quartz
(General)

<table>
<thead>
<tr>
<th>Property</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Molar mass</td>
<td>0.060 kg/mol</td>
</tr>
<tr>
<td>Density at STP</td>
<td>2650 kg/m³</td>
</tr>
<tr>
<td>Melting point</td>
<td>1850–2000 K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>2500–2750 K</td>
</tr>
<tr>
<td>Refractive index</td>
<td>1.53–1.56</td>
</tr>
</tbody>
</table>

 FIGURE Q.6  Illustration of the setup employed for the two measurements required to find the Quality Index of a linac x-ray photon beam.
The mineral quartz (from German ‘Quarz’) is the most abundant mineral in the Earth’s continental crust. Quartz is a rhombohedral crystal consisting of a lattice of silica (silicon dioxide, SiO$_2$) tetrahedra (Figure Q.7). Naturally occurring quartz crystals are normally twinned or distorted, lacking macrocrystalline structure. Pure quartz is usually colourless or white, although coloured forms exist, such as rose, amethyst and smoky. The various types of quartz depend on the specific macro and microcrystalline structure, though generally the more transparent varieties tend to be macrocrystalline in structure.

Quartz is widely used in the production of concrete and glass, and it is a major ore of silicon, which is used in integrated circuits (‘chips’). Quartz exhibits the phenomenon of piezoelectricity, meaning that it generates an electric potential when under mechanical stress. This property led to the use of quartz in specific applications such as phonograph pickups and crystal oscillators in electric circuits of watches and computers. The quartz clock utilises its natural resonant frequency to accurately mark time. It is used in devices for accurate measurements, including microbalances, strain gauges and thin-film thickness monitors. These measurements are accomplished through mechanical loading which changes the resonant frequency of the quartz crystal oscillator. Piezoelectric sensors function by producing a force on the opposing faces of a sensing element due to the change in its physical dimension. These sensors are often used as extremely sensitive microphones, such as those used in industrial non-destructive testing (NDT). They often act as both a transmitter and receiver of sound because the piezoelectric property is reversible.

**Medical Applications:** Quartz can be used in piezoelectric sensors of ultrasonic transducers for medical imaging, although the ceramic lead-zirconate-titanate (PZT) is more commonly used. Ultrasound transducers are also used in physiotherapy, lithotripsy to obliterate kidney stones and HIFU (high intensity focused ultrasound) for destruction of pathogenic tissue.

**Related Articles:** Ceramics, HIFU, Piezoelectric crystal in ultrasonography, Potential difference, PZT, Ultrasound

**Quench (quenching)**

**Magnetic Resonance** A quench is the sudden loss of superconductivity of the magnet coils due to a local temperature increase in the magnet above the critical temperature of the material used for the coils. The material most commonly used for the coil construction is niobium titanium (Nb-Ti) which has a critical temperature of 9 K. To keep constantly the coils at a temperature lower than 9 K, the Nb-Ti filaments are immersed in liquid helium at a temperature of 4.2 K. To avoid the possibility of local transitions to the resistive state the filaments of Nb-Ti are enclosed in a matrix of copper, good thermal conductor, to ensure a stable temperature state. During a quench the energy of the magnetic field is converted into heat. The energy is equal to $1/2 LI^2$ where $L$ is the inductance and $I$ is the coil current. After releasing a quench, the magnetic field strength drops to approx. 20 mT within approx. 20s. The helium evaporates rapidly producing a huge amount of helium gas that has the potential to displace all the oxygen in the magnet room if it is not carried away as the volume ratio of the gaseous to liquid helium is 730. The rapid escaping of helium displaces the oxygen in the room and there is a real danger of asphyxia if the lighter than air gas displaces all the air. The oxygen concentration in the room can be checked via an oxygen metre and in case of low oxygen concentration they automatically command a dedicated forced ventilation of the examination room. In case the exhaust line fails in part or in full, the heating and air-conditioning system is not capable of guaranteeing sufficient air exchange. Heavy fog formation toward the ceiling of the examination room impairs visibility. Also, the pressure in the examination room will rise. Depending on the type of defect, e.g. large leaks, hazards such as acute hypothermia or frostbite are present. When helium escapes, the condensation process at pipes or at the magnet may lead to local oxygenation with an increase in fire hazards in the vicinity of these components.

**Quenching**

**Nuclear Medicine** Quenching refers to any process that interferes with the counting performance of counting system. In many liquid scintillators dissolved oxygen works as a quenching agent leading to a reduction in fluorescence efficiency. Liquid scintillators are therefore sealed in a closed volume from which most of the oxygen has been extracted. There are two types of quenching; colour quenching and chemical quenching. Colour quenching means that the added sample induces a change in the optical properties of the solvent so that the scintillation light is unable to penetrate the solvent. Chemical quenching refers to sample-induced chemical changes that interfere with the scintillation energy transfer process.

Quenching used in Geiger–Müller (GM) counters is necessary to limit the amount of detector dead time. The quenching is accomplished with use of a small quantity of alcohol vapour in the gas.

**Related Articles:** Liquid scintillation (LS) counting, Chemical quenching, Geiger–Müller (GM) counters

**QUIPSS – QUIPSS II – Q2TIPS**

**Magnetic Resonance** The arterial spin labelling (ASL) technique QUIPSS (quantitative imaging of perfusion using a single subtraction), the modified version QUIPSS II (Wong et al., 1998) and Q2TIPS (QUIPSS II with thin-slice TI, periodic saturation) (Luh et al., 1999) are all ASL saturation techniques which are less sensitive to transit delays, and all provide quantitative perfusion images based on measurements at a single inversion time. QUIPSS and QUIPSS II differ in the application of the saturation pulses. In QUIPSS, the saturation is applied in the imaging slice after a delay time (TI1), and the signal in the difference image is based on the blood that arrives between saturation and the readout pulse. In QUIPSS II, the saturation is applied in the inversion slab instead, and only blood that has left the inversion slab before saturation contributes to the signal in the difference image. Q2TIPS is a modified QUIPSS II saturation technique, where the QUIPSS II saturation pulse is replaced by a periodic train of thin-slice saturation pulses at the distal end of the inversion slab. This modification is introduced to minimise the errors related to the incomplete saturation of the spins of inverted blood occurring in a thick saturation pulse. Figure Q.8 illustrates the Q2TIPS pulse sequence and the corresponding slice positions: Two in-plane presaturation pulses are applied to the imaging slices for further reduction of signal from...
the static tissue. The presaturation is followed by a 180° inversion, usually a pulse applied to a 10 cm tagged region. After a delay TI, the train of thin-slice saturation pulses is applied at the distal end of the tagged region. Finally, the images are acquired by a fast read-out sequence (e.g. EPI, spiral reconstruction, HASTE or 3D GRASE) in a region distal (superior) to the inversion slab and with a 1 cm gap between the inversion slab and the imaging region. The control image is acquired according to either the EPISTAR or the PICORE preparation method.

**Related Articles:** Perfusion imaging, Arterial spin labelling, EPISTAR, FAIR, PICORE

R (Röentgen)

(Radiation Protection) The unit of Exposure is the Röentgen, R. It is named after Wilhelm Röentgen, German Physicist, discoverer of x-rays (1845–1923). 1 R is equivalent to 1 C of electrical charge of one sign, liberated in 1 kg of matter.

**Related Articles:** Röentgen (R)

Rad (radiation absorbed dose)

(Radiation Protection) See Radiation absorbed dose (Rad)

Radiation

(Radiation Protection) Radiation is a process by which energy is transferred from one point in space to another. It usually refers to the transfer of energy by electromagnetic wave or photon. The higher frequency/shorter wavelength end of the electromagnetic radiation spectrum has enough energy to break biological molecular bonds and/or to eject electrons from their atomic orbits – that is causing ionisation – this radiation therefore being called ionising radiation.

**Related Articles:** Radiation, Gamma rays, Quanta, Ionisation, Ionising radiation, Non-ionising radiation

Radiation absorbed dose (Rad)

(Radiation Protection; General) Originally defined in the first recommendations of the ICRP in 1958 (Publication 1), this was the unit of absorbed dose until superseded by the Gray. 1 Gy = 100 rad. The unit is still used, particularly in the United States.

**Related Article:** Absorbed dose

Radiation, alpha

(Radiation Protection) Alpha radiation is composed of alpha particles emitted during radioactive decay of nuclides that decay by alpha decay.

An alpha particle (α) is a helium nucleus and consists of two neutrons and two protons. It therefore has a total mass of four. Alpha particles are typically emitted with energies between 3 and 7 MeV.

Alpha radiation has a very short range due to the mass and double charge of the alpha particles. This causes a very high ionisation density and LET.

Alpha decay normally occurs when very large nuclei decay – an example is the decay of radium 226:

\[
\frac{226}{88}\text{Ra} \rightarrow \frac{222}{86}\text{Rn} + \alpha
\]

**Abbreviation:** LET = Linear energy transfer.

**Related Articles:** Alpha particle, Alpha particle emitter, Radioactive decay

Radiation, beta

(Radiation Protection) See Beta radiation

Radiation biology

(Radiation Protection) Both ionising and non-ionising radiation may cause damage to cells, leading to biological effects. These effects are dependent on the type and energy of the incident radiation and are described under Bioeffects. The science studying such radiation damage and biological effects on the human body is known as radiation biology, or radiobiology, leading to the description of radiobiological models to predict the harm caused by exposure to ionising radiation.

**Related Articles:** Bioeffects, Radiation damage, Radiobiological models

Radiation, bremsstrahlung

(Diagnostic Radiology)

See Bremsstrahlung

Radiation, Čerenkov

(Radiation Protection) See Čerenkov radiation

Radiation damage

(Radiation Protection) Radiation damage may be classified into two types of effects:

1. Direct effects
2. Indirect effects

Direct effects tend to be caused by charged particle ionising radiation such as protons, electrons and alpha-particles. The radiation is absorbed directly by the critical target, especially DNA in the nucleus of the cell. Direct damage to the DNA in the form of strand breaks may mean that cell functions are impaired and cell death may occur.

Indirect effects tend to be caused by photonic ionising radiation such as gamma rays and x-rays interacting with water molecules in the cytoplasm of the cell. The physical interaction is followed by chemical reactive stages and a subsequent biological effect on the cell. For more details, see the article on Radiolysis.

**Repair of Radiation Damage:** When cells are exposed to ionising radiation, this damage does not necessarily lead to adverse effects or bioeffects. Cells have proteins and enzymes whose main function is to act as part of the mechanisms to repair damage to DNA. Ordinarily these mechanisms work very well to repair radiation damage.

**Classification of Radiation Damage:** The damage caused by radiation may be described in terms of the effect it has upon the individual (somatic damage), the reproductive cells (genetic damage) and the unborn foetus (teratogenic damage).

**Somatic Damage:** There is a latent period associated with somatic damage. The latent period is the length of time between irradiation and tumour development and varies depending on the type of tumour that develops and the age of the exposed person. There are two risk models that may be applied to describe the development of cancer post-latency: additive (absolute) and multiplicative (relative).

**Genetic Damage:** Genetic damage is caused by damage to the DNA in the gonads due to radiation exposure. Damage to the DNA leads to an increase in the probability of hereditary diseases; these may be dominant (first generation), recessive (later generations) or x-linked (first generation).
**Teratogenic Damage**: Effects have been observed in children who were in utero during severe radiation events such as the Japanese atomic bomb. These observations suggest that damage to the foetus caused by radiation may lead to effects such as mental retardation.

**Related Articles**: Radiolysis, Bioeffects, Deterministic effects, Stochastic effects

**Radiation detection systems** *(Radiation Protection)* Radiation detection systems are used for the detection of radioactivity, for measuring dose and for identification of the type and the energy of the radiation.

The simplest radiation detection system, for detecting radioactivity, consists of a Geiger–Müller counter or semiconductor detector with a power supply and a number of counts or counts/s indicator.

For example, the detection systems applied x-ray equipment in diagnostic radiology with semiconductor detector measures an entrance dose, dose indicator (for digital radiography), and for computer tomography, a dosimetre with a CT chamber and CT adapter, measures the computed tomography dose index (CTDI$_{100}$) and weighted computed tomography dose index (CTDI$_w$).

The detection system for identification of the type and energy of radiation should consist of a detector not only sensitive to the type of radiation ($\alpha$, $\beta$, gamma or X) but also able to produce output pulses with height proportional to the energy of the radiation. The detection systems generally used are proportional gas-filled detectors, scintillation and liquid scintillation counters and germanium detectors.

Pulse processing and shaping must be done with appropriate electronics consisting of preamplifier, amplifier and pulse height analysis system using single- or multichannel analysers.

The energy spectrum is represented as a function of the measured intensity of radiation against its energy (corresponding to the channel number).

**Abbreviations**: CTDI$_{100}$ = Computed tomography dose index and CTDI$_w$ = Weighted computed tomography dose index.

**Related Articles**: Geiger–Müller (GM) counters, Germanium detectors, Liquid scintillation (LS) counting, Proportional counter, Pulse-height analysers (PHAs) for radiation detectors, Scintillation detector


**Radiation dose** *(Radiation Protection)* Radiation dose is defined as the energy absorbed per unit of mass. When dose is measured using an ionisation chamber to calculate radiation exposure in Roentgens (R), then the exposure is calculated by dividing the charge in Coulombs (C) detected by the ionisation chamber, divided by the mass of air (kg) within the chamber, that is C/kg.

The interactions that occur when an x-ray photon passes through an ionisation chamber lead to the absorption of kinetic energy (measured in Joules) within the air molecules. This energy absorption is termed absorbed dose. Absorbed radiation dose is measured in Gray (1 R = 8.7 × 10$^{-3}$ Gy), 1 Gy = 1 J/kg.

Absorbed dose may be converted into equivalent dose by applying factors to take into account the type of incident radiation. These radiation weighting factors are defined by the International Commission on Radiological Protection (ICRP). Equivalent dose is measured in Sieverts (Sv).

Equivalent dose, in turn, may be converted to an effective dose by applying factors to take into account the tissue that the radiation is incident upon. These tissue weighting factors are also defined by the ICRP. Effective dose is also measured in Sieverts.

**Related Articles**: Ionisation chamber, Absorbed dose, Equivalent dose, International commission on radiation protection, Effective dose

**Radiation dosimetry** *(Radiation Protection)* See Dosimetry

**Radiation, electromagnetic** *(Radiation Protection)* Electromagnetic (EM) radiation is the propagation of energy by simultaneous vibration of electric and magnetic fields. EM radiation is characterised by its wavelength (10$^{-13}$–10$^3$ m). Radiation wavelength, frequency and quantum energy are used to define the electromagnetic spectrum, different ranges of the spectrum are recognized by different names: radio waves, infra-red, visible light, ultraviolet, x-rays and gamma rays.

The different EM radiation ranges in the EM spectrum are further grouped into ionising and non-ionising radiation. About 34eV of energy is required to produce an ion pair by the process of ionisation. A quantum of x-rays or gamma rays will always possess at least 34eV of energy and are referred to as ionising radiation. However, a quantum of radio waves, infra-red, visible light or ultraviolet rays will possess less than 34eV and these wavelength ranges are referred to as non-ionising radiation. Ionising radiation is used more frequently than non-ionising radiation for medical purposes.

The properties of the electromagnetic spectrum and its parts, ionisation, ionising and non-ionising radiation are discussed in greater detail elsewhere.

**Related Articles**: Radiation, Gamma rays, Quanta, Ionisation, Ionising radiation, Non-ionising radiation, Ultraviolet radiation (UV), Infrared radiation (IR)


**Radiation exposure** *(Radiation Protection)* See Exposure

**Radiation field** *(Radiation Protection)* The radiation field is the area irradiated by the incident radiation beam and is defined at a set distance from the radiation source. The intensity of the radiation may vary across the field and may be altered by the use of filtration. The size and shape of the radiation field is defined and altered by the use of collimation.

**Radiation force** *(Ultrasound)* The radiation force is defined as time-average force acting on a body in a sound field and caused by the sound field. More generally it is the time-average force in a sound field, appearing at the boundary surface between two media of different acoustical properties [1]. Unit: Newton, N.

Radiation force is measured by an ultrasound force balance and the relationship between the force F and acoustic power W can be expressed as $F = hW/c$, where $c$ is the speed of sound and $h$ is a constant depending on the geometry of the used target.
**Radiation force balance**

(Ultrasound) See *Force balance*

**Radiation, gamma**

(Radiation Protection) See *Gamma radiation*

**Radiation hazard**

(Radiation Protection) Hazard, or *health hazard*, is defined as anything that has the capacity to cause harm to a human being. If the hazard relates to radiation exposure, it may be called a radiation hazard.

**Related Articles:** Hazard, Risk assessment

**Radiation, infrared**

(Radiation Protection) See *Infrared radiation*

**Radiation, ionising**

(Radiation Protection) See *Ionising radiation*

**Radiation isocentre**

(Radiotherapy) This is the point (rather small spherical volume) in space about which all the radiation beams from different gantry angles overlap, and should be in good agreement with the mechanical isocentre. Prior to performing this test the photon collimator symmetry should be checked for two collimator angles 180° apart.

To find the radiation isocentre place a piece of radiographic film vertically in-between some sheets of Perspex on the treatment couch such that the centre of the film is roughly at the isocentre (use the in-room lasers for this). The jaws should be closed to a very small field size (e.g. 1 × 1 cm² or smaller), and sufficient MUs delivered for each field to give a readable optical density on the film. Without moving the film irradiate at a range of different gantry angles (four or five exposures from the different quadrants of gantry angle) taking care to avoid any overlap between angles. The processed film will have a series of lines crossing it and is commonly referred to as a ‘star-shot’ film (an example is given below in Figure R.1). Mark the centre line of each beam on the film. The point (or rather small region) where the central axis of all the fields overlap is the radiation isocentre, and should ideally be no more than 1 mm in width, that is a radius less than 0.5 mm.

**Related Articles:** Isocentre, Mechanical isocentre

**Radiation monitoring**

(Radiation Protection) In addition to what is described under *Dose monitoring*, radiation monitoring requires a more general approach including: identification of hazard and risk, the availability of adequate instrumentation and of know-how and trained personnel, the identification of roles and responsibilities, the formulation and implementation of programs and the evaluation of program effectiveness.

**Related Articles:** Dose monitoring

**Radiation, nuclear**

(Radiation Protection) Radiation describes any process in which energy emitted by a source in the form of particles or electromagnetic waves travels through a medium or through space. It also refers to the energy itself. ‘Nuclear radiation’ is when the energy emitted is the result of a nuclear process, such as radioactive decay or by the nuclear transformation of matter by bombardment with particles.

**Related Articles:** Radioactive decay, Radiation

**Radiation, particle**

(Radiation Protection) See *Particle radiation*

**Radiation, penetrating**

(Radiation Protection) See *Penetrating radiation*

**Radiation, positron**

(Radiation Protection) See *Positron decay*

**Radiation pressure**

(Ultrasound) In acoustics, the radiation pressure is normally defined to be the pressure force that a beam of sound inflicts on a target, or at the interface between two media. The force exerted on the target can be plane, conical or spherical and is related to the ultrasound intensity or total power. There are two main models for describing the theoretical background of the force.

The *Rayleigh radiation pressure* is known to be the force per unit area on an absorber. The theory is based on a model representing oscillations in a sealed tube, where one end of the tube is a sound source and the other end the absorber.

If the source and absorber are placed in a fluid, unlimited in all directions, the resulting force on the absorber will differ. This force is known as the *Langevin radiation pressure*. The magnitude of the radiation pressure is proportional to the kinetic energy density in the sound wave.

**Related Articles:** Radiation power, Force balance

**Radiation, primary**

(Radiation Protection) See *Primary radiation*

**Radiation protection**

(General) Radiation protection is a rather general term. Radiation is a form of energy that has always been around in nature and will
always influence human life (natural radiation). In addition to the general presence in the environment, humans have found many forms of specific use of ionising radiation (man made), for example energy production, industry and medicine (not to talk about atomic bombs). When the effects of ionisation and non-ionisation radiation were studied, a necessity to put in place regulations and control systems arose as a consequence, in order to limit the negative effects of radiation and optimise its use.

In order to assess the impact of radiation exposure properly, it is essential to introduce appropriate quantities and units, for the various kinds of radiation, which can then be used for the quantification of exposures (see Dosimetry). In principle, the main aim of radiation protection is the control of radiation exposure, while radiation dosimetry deals primarily with the measurements of relevant radiation quantities, in particular doses.

In summary, in relation to the different kinds of radiation, it is essential to know the relevant radiation characteristics definitions, quantities, interpretations and effects. The study of short and long-term effects on the environments and human beings along with cost-benefit evaluation (exposure of population, radiation workers and patients) are the basis for the implementation of an appropriate radiation protection system.

**Radiation protection adviser (RPA)**

*(Radiation Protection)* In the United Kingdom, the Ionising Radiations Regulations 1999 (IRR99) requires employers to appoint and consult a radiation protection adviser (RPA) if the work they undertake involves the use of ionising radiation (whether electrically produced in x-ray tubes, etc. or from work with radioactive substances). There are a few exceptions to this requirement based on very-low-hazard work – for example, radioactive substances with very low specific activity, or x-ray generating equipment operating below 30kV and with a dose rate no greater than 1μSv h at 1 m from any surface of the equipment.

Although the employer may wish to consult the RPA about any aspect of the Regulations, it is compulsory for the employer to consult the RPA on the issues listed in Schedule 5 of the Regulations, which are reproduced here:

1. The implementation of requirements as to controlled and supervised areas.
2. The prior examination of plans for installations and the acceptance into service of new or modified sources in relation to any engineering controls, design features, safety features and warning devices provided to restrict exposure to ionising radiation.
3. The regular calibration of equipment provided for monitoring levels of ionising radiations and the regular checking that such equipment is serviceable and correctly used.
4. The periodic examination and testing of engineering controls, design features, safety features and warning devices and regular checking of systems of work provided to restrict exposure to ionising radiation.

The appointment of an RPA must be made in writing by the employer. The employer is expected to provide the RPA with adequate information and facilities to enable him to undertake his/her work.

The qualifications and experience required for an individual to become an RPA is described by the health and safety executive (HSE). A person wishing to gaining certification as a RPA needs to demonstrate to an HSE recognised certification body that he/she possesses the knowledge and experience to enable him/her to undertake the work satisfactorily. Once a certificate is issued to the individual, it must be renewed least every 5 years, involving the person demonstrating continuing professional development.

**Radiation protection officer (RPO)**

*(Radiation Protection)* The radiation protection officer (RPO) is a professional of proven competence in radiation protection matters relevant for a given type of practice, who is appointed by the employer or licensee in order to ensure that the requirements of the national standards are applied.

National radiation protection laws and regulations normally include role, duties and responsibilities of the RPO and/or qualified expert (see also Qualified expert) or other designated persons with specific responsibilities in radiation protection. Different national regulatory authorities and national laws and regulations can use different terminologies.

**Related Articles:** Qualified expert, Regulatory authority


**Radiation protection supervisor (RPS)**

*(Radiation Protection)* In the United Kingdom, the implementation of the Basic Safety Standards Directive (96/29/Euratom) is achieved by The Ionising Radiations Regulations 1999 (IRR99). Regulation 17(4) of the IRR99 requires a radiation protection supervisor (RPS) to be appointed for any designated radiation work area that has been made subject to Local Rules – a document setting out procedures for ensuring that no one entering the area receives an unacceptably high radiation dose. The role of the RPS is to ensure anyone working in the area does so in compliance with the Local Rules. However, legal responsibility for compliance remains with the employer.

The RPS is appointed by the employer and should be named in the Local Rules. There may be more than one RPS for an area and an RPS may be responsible for more than one area.

The Approved Code of Practice suggests that a suitable RPS should:

- Know and understand the requirements of the Regulations and Local Rules relevant to the work with ionising radiation
- Command sufficient authority from the people doing the work to allow them to supervise the radiation protection aspects of that work
- Understand the necessary precautions to be taken and the extent to which these precautions will restrict exposures
- Know what to do in an emergency

In general the RPS will be an employee of the employer undertaking the work, in a line management position and closely involved with the work being done. However the RPS does not have to be present and directly supervising the work being done. The RPS should receive appropriate training that reflects the complexity of the work undertaken, so they can fulfil their task.

In practice the RPS often acts as the focus point for radiation protection issues that may arise in the work area and will need to liaise with the employer and radiation protection advisor to ensure compliance with regulations and safe working practices are maintained.

**Related Articles:** Local rules, Radiation protection adviser
Radiation quality

(Radiation Protection; General) This is the quantity given to each type of ionising radiation to quantify the relative effectiveness in causing damage when the radiation interacts with cells. This is also known as the radiation weighting factor.

Related Articles: Radiation weighting factor

Radiation safety instrument

(Nuclear Medicine) In each laboratory where one is using ionising radiation, one needs to have radiation safety instruments, designed to monitor the radiation environment in each laboratory. There are several types of instruments.

A dose-rate monitor gives continuous indication of the radiation intensity at one or more points in the laboratory where the radiation intensity may change owing to the work carried out in the laboratory.

An area monitor is a dose-rate monitor that is usually permanently installed in a nuclear medicine laboratory. Its purpose is to check the radiation intensity in such laboratories as hot lab, accelerator rooms, storage rooms, etc.

A laboratory bench monitor is a semi portable dose-rate monitor that is ordinary kept in the laboratory work area. This is especially important when working with open radiation sources as in a radiochemistry laboratory. It is also useful for making rough radioassays, for checking the worker’s hands for contamination, as well as the laboratory working area for radioactive spill.

A personnel monitor or personal radiation alarm is a dose-rate monitor to be carried in a pocket.

A survey meter is a compact, portable radiation intensity measuring instrument. It can be used for monitoring adequate shielding, locating areas that have been contaminated, etc.


Radiation safety officer (RSO)

(Radiation Protection) The term radiation safety officer is used specifically in the United States of America. The RSO is the person within an organization responsible for the safe use of radiation as well as regulatory compliance. An organisation licensed by the Nuclear Regulatory Commission to use radioactive materials must designate a radiation safety officer in writing.

The requirements for a radiation safety officer (RSO) vary with the type of license and types of materials used. Requirements are laid down in the regulations of the US Nuclear Regulatory Commission (NRC) and Regulatory guides (NUREG), reports and brochures from the US Nuclear Regulatory Commission (USNRC).

The term radiation safety officer is sometimes used in other countries for a person within an organization who has some level of responsibility for radiation protection. The duties may be similar to those of a radiation protection supervisor or of a broader responsibility comparable to a radiation protection officer. National legislation and guidance should be checked to ensure that the correct national terminology is used.

Related Articles: Nuclear regulatory commission (NRC), Radiation protection officer (RPO), Radiation protection supervisor (RPS)


Radiation, scattered

(Radiation Protection) Scattered radiation refers to all photons and charged particles resultant from scattering interaction between an incident photon or particle of ionising radiation and the medium.

Scattered radiation may travel in any direction away from the site of interaction. Therefore further ionisations and energy deposition in the medium may occur outside the primary radiation beam. Interactions between scattered radiation and the medium are important for radiation protection purposes – they are important in radiotherapy treatment planning when attempting to deliver the prescribed dose to the target volume (tumour) whilst minimising the peripheral dose to surrounding healthy tissues due to such scattered radiation.

Scattered radiation may also exit the medium altogether. Such transmitted scattered radiation is important in diagnostic radiology where it may be detected by the imaging receptor, contributing to the ‘fogging’ of the x-ray image. Scattered radiation is minimised in diagnostic radiology by the introduction of anti-scatter bucky grids between the patient and the image receptor.

Related Articles: Secondary radiation, Secondary electron, Peripheral dose, Bucky grids

Radiation, secondary ionising

(Radiation Protection) Secondary ionising radiation refers to all photons and charged particles resultant from an interaction between an incident photon or particle of ionising radiation and the medium, which still have enough energy to cause further ionisations. Therefore secondary ionising radiation does not include any secondary photons or charged particles which are non-ionising, for example ultraviolet radiation.

Secondary ionising radiation may travel in any direction away from the site of interaction. Therefore further ionisations and energy deposition in the medium may occur outside the primary radiation beam. Secondary interactions are important for radiation protection purposes – they are important in radiotherapy treatment planning when attempting to deliver the prescribed dose to the target volume (tumour) whilst minimising the peripheral dose to surrounding healthy tissues.

Related Articles: Secondary radiation, Secondary electron, Peripheral dose

Radiation shielding

(Radiation Protection) Shielding is one of the methods, together with reducing time and increasing distance (see also Time–distance–shielding), to reduce radiation exposure. When the optimisation of time and distance is not sufficient, there is a need to provide shielding barriers. The shielding barriers protect staff, patients (when not being examined/treated), visitors and public, person working adjacent to or nearby. The intensity of radiation falls exponentially going through the shielding.

In general the basic information needed to calculate a barrier are: (1) equipment type, (2) workload (W), (3) target dose (D), (4) use factor and direction of primary beam (U), (5) distance to the area of interest (d), (6) occupancy of area to be shielded (T), (7) limit values in area to be shielded (P). The evaluations are made on the basis of possible working conditions, in such a way that, there should be a reasonable safety margin. Under shielding is worse than over shielding, but shielding is usually expensive, therefore it should be reasonably evaluated. There are many computer programs available to calculate the thickness of various materials. The 10th value layer thickness (TVL) for different material is usually given.
The problems related to shielding in diagnostic radiology and nuclear medicine are clearly different from those in radiotherapy where a potentially lethal dose is given to patients.

The choice of material depends very much on the setting and kind of radiation. There are also situations when improper shielding can make the situation worse.

Typical shielding situations are as follows:

1. Low-energy gamma and x-ray can easily be shielded by lead or by any other material if used in sufficient amount.
2. For high energy (more than 500keV) gamma and x-rays, concrete (high-density concrete) is recommended, as it is cheaper and self-supporting.
3. Electrons are shielded appropriately when photons are accounted for.
4. Neutrons from high-energy linacs need special consideration.

**Lead:** Lead is a high-density (11.3 g/cm³) material with a high atomic number, requires small space and is a good shielding for low-energy x-rays. Wooden panels with lead inside can be added to already existing walls. It is relatively expensive and difficult to work with as it can easily slide.

**Concrete:** Concrete is the best material when used during construction. The density can vary from 1.8 to 3.7 g/cm³ for the high-density kind, and has to be tested. It is a cheap material.

**Brick:** The density of bricks can vary very much depending from the shape, for example solid or with holes.

The composition is mostly clay-earth with a density of 1.6 g/cm³.

**Iron/Steel:** The density is circa 7.8 g/cm³ and its contribution has to be considered. The contribution from these materials in the equipment must be considered.

**Ledite:** Ledite has a density of about 4 g/cm³ and is a mixture of lead in concrete bricks. The availability in bricks makes it possible to use this material also as additional self-supporting shield.

Different materials can be used; the thickness required varies very much. Just to give an example, for energy in the 500kVp spectrum, the 10th value layer thickness, TLV, is 1.19 cm for lead and 11.7 cm for concrete with a density circa 2.4 g/cm³.

The presence of corners might present particular problems and it is important to create overlapping of the protective material. The same applies for the frames around windows and doors.

The simple calculation of shielding is not enough; it is compulsory to verify this with measurements at the barriers at the moment of their construction and also regularly, with adequate measuring equipments and procedures, as part of the departmental monitoring.

**Related Articles:** Time–distance–shielding (TDS) rules, Lead content


**Radiation, ultraviolet**

(Radiation Protection) See Ultraviolet radiation

**Radiation weighting factor**

(Radiation Protection) Different types of ionising radiation (e.g. x-rays, alpha particles, beta particles and gamma rays) each have a different relative biological ability to induce damage to biological molecules, including DNA, dependent on the density of ionising events caused by the radiation. This ability, or effectiveness to cause damage, is used in the conversion of the absorbed dose to a tissue or organ to the equivalent dose to that tissue, and can be represented numerically in the form of a radiation weighting factor. For more information, see the articles on Equivalent dose and Relative biological effectiveness.

**Related Articles:** Equivalent dose, Relative biological effectiveness

**Radio waves**

(General) See Electromagnetic energy spectrum

**Radioactive decay**

(General) Nuclides that do not have a stable combination of neutrons and protons undergo spontaneous radioactive decay. This transformation involves either the expulsion of a charged particle or the capture of an electron by the nucleus, with the result that the nucleus is either stable or more stable (lower-energy state). This process alters the balance of neutrons to protons and results in an atom of a different element. The nucleus may be left in an excited state by the transformation and will in most cases transition to lower-energy (ground) state by emission of one or more gamma rays; in some cases, however, internal conversion (IC) occurs with the emission of one or more electrons from the atom (Figure R.2).

**Abbreviation:** IC = Internal conversion

**Related Articles:** Isotope, Radioisotope, Gamma ray

**Radioactive materials**

(Radiation Protection) Radioactive materials are composed of, or contain, one or more radioactive substances in sufficient quantities to constitute a hazard to health. The radioactive substance, or substances, may be naturally occurring or artificially produced by nuclear fission or by bombardment by neutrons or ionising radiation.

National authorities define the levels of radioactivity above which a material is classified as being a radioactive material. Also some national authorities will exclude radioactive waste from the definition of radioactive waste.

Radioactive materials are often classified in different types such as unsealed radioactive sources and sealed sources.

**Related Articles:** Radioactive substance, Radioactive waste

**Radioactive series**

(Nuclear Medicine) See Parent–daughter decay

**Radioactive sources**

(Radiation Protection) A radioactive source is any object or material which emits ionising radiation because it contains a radioactive substance. Radioactive sources are manufactured or produced for a specific purpose and are divided into two classes – sealed sources that utilise radioactive materials that are firmly contained or bound

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**Related Articles:** Equivalent dose, Relative biological effectiveness

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**Related Articles:** Equivalent dose, Relative biological effectiveness

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(Nuclear Medicine) See Parent–daughter decay

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**Related Articles:** Equivalent dose, Relative biological effectiveness

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Radioactive tracer

(Nuclear Medicine) This is the common name for tracer compounds labelled with a radioisotope, for example 99mTc, 125I and 18F. In nuclear medicine radioactive tracer is often referred to as ‘tracer’. A more elaborate explanation on tracers can be found in the separate article Tracer.

Related Article: Tracer

Radioactive waste

(Radiation Protection) Radioactive waste is radioactive material for which no further use is foreseen. Radioactive waste is usually divided into different categories such as high-level waste (from the reprocessing of spent nuclear fuel), intermediate-level waste and low-level waste (generally in the form of radioactively contaminated industrial, medical or research waste). Special categories are also used such as spent nuclear fuel from reactors and naturally occurring radioactive materials. The detailed specification of these categories is laid down by national regulatory authorities.

The Basic Safety Standard of the International Atomic Energy Agency (1996) defines radioactive waste as ‘Material, whatever its physical form, remaining from practices or interventions and for which no further use is foreseen (i) that contains or is contaminated with radioactive substances and has an activity or activity concentration higher than the level of clearance from regulatory requirements, and (ii) exposure to which is not excluded from the Standards’.

Related Articles: Radioactive material, IAEA


Radiobiology

(Radiotherapy) Radiation biology or radiobiology is the study of the interaction of radiation with biological systems, ranging from cells to biological tissues and whole organisms. It brings together the disciplines of physics, biology and chemistry and covers the effects of both ionising and non-ionising radiation. Therefore it has relevance in many areas of medical physics, including diagnostic radiology, radiation protection, MRI, ultrasound, nuclear medicine and radiotherapy.

Radiation can be both of benefit and harm to the population, for example it can both cure and cause cancer. Due to the large amount of radiobiological data collected since x-rays were discovered in 1895 by Wilhelm Roentgen, the balance of benefit and detriment can be estimated with a good degree of accuracy. Further information on the biological effects of radiation from a radiation protection point of view can be found in the article on Bioeffects.

In radiotherapy, radiobiology has aided in

1. The identification of the mechanisms and processes underlying the response of tumours and normal tissue to irradiation (for further details see the article on The 5Rs of radiobiology)
2. The development of new treatment strategies such as hyper- and hypofractionation, the use of radiosensitisers and high-LET radiation, all intended to maximise the therapeutic effect of radiotherapy
3. The selection of treatment schedules using dose–response models, for example linear quadratic model based formulae, to calculate the effect of changes in fractionation or dose rate

While the linear quadratic model and its associated formulae have been relatively successful in providing a quantitative evaluation of the clinical situation, it is not without its limitations (see the article on Linear quadratic (LQ) model and clinical trials are still essential for the final selection of treatment protocols.

Abbreviations: LET = Linear energy transfer and MRI = Magnetic resonance imaging.

Related Articles: Radiation biology, Bioeffects, The 5Rs of radiobiology, Hyperfractionation, Hypofractionation, Radiosensitisers, Therapeutic effect, Relative biological effectiveness, Dose–response models, Linear quadratic (LQ) model, Fractionation, Dose rate dependence


Radiobiological models

(Radiation Protection) Radiobiological models refers to a series of models of varied complexity designed to describe the interaction of ionising radiation with cellular DNA and other biological molecules, and to predict the damage that will be caused by a given radiation dose, whether delivered acutely at one time, or chronically over a period of time.

The models are based on an understanding of the interaction of ionising radiation with biological molecules, which can be classified as either ‘direct action’ or ‘indirect action’.

Direct action interactions are those where the incident photon or charged particle directly ionises an atom in the DNA molecule, leading to molecular bond dissociation and a strand break.

Indirect action involves the physical interaction of the radiation with water molecules in the cytoplasm of the cell, leading to a chemical reaction which in turn causes biological damage to the DNA and the cell. For more information on this indirect action, see the article on Radiolysis.

When there is damage to DNA as a result of the direct or indirect action of ionising radiation, the likelihood is that the damage will be repaired. However, if the damage is not repaired, then that damage may be expressed as either prompt deterministic effects due to cell killing or late stochastic effects due to cell mutation.

All current radiobiological models are based on the interpretation of data from exposure of persons to higher doses of radiation – the survivors of the atomic bombs at Hiroshima and Nagasaki, and from radiotherapy treatments amongst others. This leads to a ‘gold standard’ of known radiation damage and response at doses from approximately 0.1–4Sv. However much
Radiochemical purity

(Nuclear Medicine) The radiochemical purity (RCP) of a radio-pharmaceutical preparation is defined as the ratio of the activity between the desired chemical form and the total activity of the preparation. The RCP value is expressed as a percentage.

Radiochemical impurities are undesirable because it may result in an unwanted distribution in the body, influencing the useful information in the study due to high background activities in areas not of primary interest, as well as unwanted absorbed doses to organs and tissues of the patient.

Free $^{99m}$TcO$_4^-$ – sodium pertechnetate present may readily be seen in the scintigraphy as increased uptake in the thyroid, salivary glands, stomach and in the bladder.

A number of sources can cause poor radiochemical purity, for example oxidation–reduction reactions, chemical changes during storage because of changes in pH, temperature or light, competing chemical reactions during labelling, preparation techniques and radiolysis.

Acceptable RCP values are listed in the package insert or in the European pharmacopeia, and must be fulfilled throughout the useful life of the radiopharmaceutical.

A number of chromatographic systems and methods are in use for testing the RCP. However, the most commonly used system is thin-layer chromatography (TLC).

**Abbreviations**: RCP = Radiochemical purity and TLC = Thin-layer chromatography.

**Related Articles**: GMP, Quality control, Radionuclide purity, Chemical purity, Biological purity


**Radiochromic film**

(Radiotherapy) This is a relatively new type of film (a.k.a Gafchromic film) that may be used for radiotherapy dosimetry. It is a coloured film which has an almost tissue-equivalent composition and produces a colour change when exposed to radiation. This colouration results from the polymerization of a dye. The polymer will absorb light and so by using a densitometer is it possible to measure the transmission of light through the film and relate this to the level of radiation exposure. A major advantage of radiochromic film compared to conventional radiographic film is that it is self-developing and therefore negates any need for developers or fixers.

Another useful feature is that the radiochromic film is grainless, so has a very high spatial resolution and can be used where there is a very steep dose gradient such as brachytherapy and stereotactic radiotherapy.

Further advantages include ease of use (i.e. no need for a dark room, film processor and chemicals, etc.), can be used for very high doses, the response is independent of dose rate and it has better energy characteristics except for very low energies (25kV).

The disadvantages are the increased cost of each film and a reduced sensitivity.

In common with radiographic film, radiochromic film is to be used as a relative dosimeter, and with appropriate care of use and conditions it should be possible to achieve a precision of better than 3%.

**Related Articles**: Radiographic film dosimetry, Gafchromic film
Radiofrequency (RF)

(Magnetic Resonance) Radiofrequency (RF) is the term used to describe any oscillating field which an oscillation frequency in the range between 3Hz and 300GHz. Most often the term is associated with electromagnetic devices, since mechanical devices cannot respond to oscillations in this range. The term ‘radiofrequency’ refers to the fact that this is the range of the electromagnetic fields used for generating and detecting radio waves.

In MRI, radio frequency pulses (RF-pulses) are used to excite the protons. In this case the pulses are generated with a frequency corresponding to the Larmor frequency of the spins in the sample. Note that this field can be generated without having any electrical component. The field is often denoted the $B_1$ field and has a magnetic field strength in the microtesla (μT) range, as determined from the duration of a 90° RF pulse.

Related Articles: RF pulse, Larmor frequency

Radiofrequency absorption

(Magnetic Resonance) See Specific absorption rate

Related Articles: Radiofrequency, Specific absorption rate

Radiofrequency heating

(Magnetic Resonance) The majority of the RF power transmitted by the RF transmitting coils for MRI or MRS is transformed into heat within the patient’s tissues as a result of resistive losses. This energy absorption is described in terms of the specific absorption rate (SAR) measured in watts per kilogram. Significant whole body heating has a major impact on the cardiovascular physiology and thermoregulatory ability and heating consequences depend on the status of the patient’s thermoregulatory system and the presence of cardiovascular disease, hypertension, diabetes, fever, old age, and obesity. The thermophysiologic responses of the patient to heating involves complex factors such as: the duration of exposure, the RF energy absorption rate, and the heat conversion in the presence of various events like blood circulation, diffusion, bulk flow. A rise in core body temperature less than 1°C generally is no cause for concern because cell death does not occur until the temperature exceeds 42°C. The acute exposure of volunteers to a 64 MHz RF for 30 min at whole body SARs of 2.7–4 W/kg or 16 min to 6 W/kg resulted in small increases of temperature (0.1°C–0.4 °C), heart rate and localized sweating and increased skin blood flow. The International Electrotechnical Commission (IEC) establishes standards concerning particular requirements for SAR limits in magnetic resonance equipment for medical diagnosis (IEC 601-2-33:1995) introducing also the concept of different operational modes. In the following table the SAR limits are reported for the different operation modes.

The imaging sequence influences the SAR as the latter increases with the square of the flip angle, the size of the patient and the duty cycle of the RF pulses. The SAR also increases with the square of the frequency and therefore depends on magnetic field strength. Thus a patient undergoing an MR imaging at 3.0T will experience nine times the SAR experienced with MR imaging at 1.0T. The temperature changes associated with RF induced heating depend on the environmental conditions that exist in and around the MR system that include ambient temperature, relative humidity and airflow. Care should be taken when imaging patients with metallic objects, which may result in an increased heating compared with the heating in normal biologic tissues and this could result in burns.

Radiofrequency screening

(Magnetic Resonance) MRI scan rooms must be screened to prevent radiofrequency signals originating from outside the room interfering with RF signal pick up from the body. MRI room designs incorporate a Faraday cage, which encloses the room with a conductive surface or mesh. The cage attenuates RF transmission into the scan room to an acceptable level (Figure R.4).

Related Article: Faraday cage

Radiograph

(Diagnostic Radiology) A radiograph is an image produced by projecting x-ray beam through an object, such as a human body, and recording the image for viewing later. Radiographs are recorded either on film or various types of electronic or digital media.

Radiographic accessories

(Diagnostic Radiology) Radiographic accessories include devices and consumable supplies that are used along with x-ray equipment to perform radiographic procedures.

Radiographic contrast

(Diagnostic Radiology) Radiographic contrast is the contrast of objects or structures in a radiograph that determines their visibility (Figure R.5).

The contrast of a specific object or area relative to its background can be measured and expressed either as the difference in density values for a radiograph recorded on film or as the ratio of brightness values for radiographs on digital displays.

<table>
<thead>
<tr>
<th>Region</th>
<th>Normal Mode Limit</th>
<th>First Controlled Operation Mode Limit</th>
<th>Second Controlled Operation Mode Limit</th>
</tr>
</thead>
<tbody>
<tr>
<td>Whole body</td>
<td>1.5 W/kg averaged over 15 min</td>
<td>4 W/kg averaged over 15 min</td>
<td>Above the first controlled mode limits</td>
</tr>
<tr>
<td>Head</td>
<td>3 W/kg averaged over 10 min</td>
<td>3 W/kg averaged over 10 min</td>
<td>Above the first controlled mode limits</td>
</tr>
<tr>
<td>Local SAR</td>
<td>8 W/kg head and torso, 12 W/kg in extremities averaged over 5 min</td>
<td>8 W/kg head and torso, 12 W/kg in extremities averaged over 5 min</td>
<td>Above the first controlled mode limits</td>
</tr>
<tr>
<td>(averaged over any 1 g of tissue)</td>
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FIGURE R.4 Copper RF screening being fitted during an MRI room build.
Radiographic film dosimetry

(Radiotherapy) Film dosimetry has been used extensively as a convenient and rapid means of measuring dose distributions of electron and photon beams and also for dynamic beams and for studying the combination of stationary beams treated sequentially. Film dosimetry is in particular important for the dosimetry of scanning electron beams where the automated dosimetry systems using diode or ionization chambers cannot be easily employed because of long accelerator beam times. Singular advantages of film as a dosimeter are its high spatial resolution, minimum perturbation of the radiation field and acquisition of 2D data simultaneously over a large area.

Radiographic films consist of silver halide crystals (dimension 0.2–2 μm), usually AgBr, embedded in gelatin and spread uniformly on a polyester base in a thin sensitive layer. The electrons freed by ionization process preferentially move into vacancies/imperfect sites in the lattice of the silver halide crystals where a silver ion is reduced to silver atom by the process

\[ \text{Ag}^+ + e^- \rightarrow \text{Ag} \]

producing a ‘latent image’. These silver specks catalyze the conversion of the whole AgBr crystal when Ag is subject to an appropriate developer solution. Then the image is fixed by dissolving the nonionised AgBr grains in a weakly acidic solution of sodium thiosulphate. The elemental silver is black and its presence determines the blackening of the film that can be quantitatively evaluated using optical densitometry. The optical density is given by

\[ \text{Optical Density} = \log_{10} \frac{I_0}{I} \]

where
- \( I_0 \) is the incident light intensity
- \( I \) is the intensity after passing through the film

The optical density is determined by the quantity of converted Ag grains and is therefore dependent on the quantity of radiation which was incident. The most common experimental set-up in relative dose measurements is to sandwich the radiographic film within a phantom of water equivalent material with the film plane parallel to the central axis of the radiation beam, taking care of the perfect alignment of the film edge with the surface of the phantom and avoiding any air gap on either side of the film, that is exerting a pressure on the phantom. To a first approximation the radiation induced optical density initially increases linearly with the exposure to radiations. If the film is exposed to photons of different energies and the optical densities lie within the linear range it is found that to a first approximation the optical density is proportional to the absorbed dose. Figure R.6 shows a set of dose–response curves in terms of optical density for the Kodak XV2 film for some beam qualities, perpendicular to the central beam axis and for a 10 × 10 cm² field size at \( d_{\text{max}} \). The films have been inserted in a water equivalent phantom for photons and polystyrene phantom for electrons.


Radiographic imaging chain

(Diagnostic Radiology) The radiographic imaging chain is the series of components and functions that produce a radiograph as illustrated in Figure R.7.

Radiographic kV control

(Diagnostic Radiology) The high-voltage generator (HVG) of the x-ray equipment has various functions, one of these being the control of kV during radiography. The main parts of the classical radiographic kV control circuit include the line-voltage compensation (through the autotransformer) and kVp selection. One important sub-function of the line-voltage compensation is the compensation of the voltage drop due to the internal resistance of the mains.

Contemporary x-ray equipment with high-frequency generator control the kV by varying the frequency of the DC–AC converter. These equipment use detectors to measure the output kV (usually by a high-voltage divider) and compare it with a value set by the radiographer. Any difference at the output of the comparator triggers the power switches of the DC–AC converter and produces current with appropriate frequency. The high-frequency current feeds the high-voltage ferrite transformer which produces the necessary output kV. This is possible because

\[ U \sim c f, \quad \text{where } c = A n \]

Here,
- \( U \) is the voltage at the secondary winding of the transformer (kV)
- \( f \) is the frequency of the current supplying the transformer (Hz)
- \( c \) is a constant depending on the cross section of the transformer core (A) and the transformer ratio \( n \) (secondary/primary windings)

![Figure R.5 Radiographic contrast and its formula. (Graphs courtesy of Sprawls Foundation, www.sprawls.org)](image)

![Figure R.6 Dose–response curves for Kodak XV-2 film in terms of optical density. (a) 120kVp, 2.5 mm Al HVL, (b) 280kVp, 1.7 mm Cu, (c) 3.5 mm Cu HVL, (d) 20 MeV electron beam, (e) Co60, (f) 6 MeV photons. (Image by Mota, H.C. et al., Phys. Med. Biol., 35(4), 565, 1990.)](image)
A good radiographic kV control would produce high voltage with less than 1% error. Usually it is measured during the quality control procedure. Error above 5% is normally an indicator of problem in the kV control circuit.

The mA control works in a similar way. In this case the comparator changes the frequency supplying the filament transformer, which varies the filament current. This changes the temperature of the cathode; hence, the flow of thermal electrons (tube current, mA).

**Related Articles:** High-voltage generator, High-frequency generator, Voltage drop, High voltage control device

**Radiographic mode**

(Diagnostic Radiology) Radiographic mode is the most often used mode of operation of an x-ray equipment. During this mode the x-ray tube is supplied with pre-set kV, mA, ms or mAs and produces a short x-ray exposure, resulting in a static (momentary) x-ray image of the organ of interest. Usually radiographic mode uses short exposures to minimize blurring due to patient organs’ movement. The time of the exposure is usually from 1 to 1000 ms. In order to produce an image with sufficient grey shades (varying from black to white) this mode applies relatively high anode current – usually from 10 to 500 mA. The kV selection depends on the x-rayed anatomical structures and the desired image contrast. Radiographic mode can use single exposure (most often for a single x-ray image) or multiple exposures (most often sequence used in x-ray angiography).

The other basic mode of operation of x-ray equipment is fluoroscopic mode (used for observing organs in their dynamics in real time). This mode uses very low anode current (usually 0.5 – 5 mA), because the fluoroscopic units have image intensifier. The length of exposure depends on the diagnosis (often more than 60 – 200 s). The kV applied in this mode also depends on the observed anatomic regions and the desired image contrast.

Although the main parts of an x-ray equipment are the same, it may have different control circuits for the different operating modes.

**Radiographic rating**

(Diagnostic Radiology) Radiographic rating is generally a clinical term referring to the process of assigning to anatomical or pathological conditions some numerical values or ‘ratings’ based on observations in radiographs.

**Radiography**

(Diagnostic Radiology) In general terms radiography is the method which uses x-rays, which pass through an object (being modulated by the absorption of its structures) and after this interact with a detector (such as film emulsion, or phosphor, etc.) this way producing a medical image.

**Radiography, digital**

(Diagnostic Radiology) See Digital radiography

**Radio-immunoconjugates**

(Radiotherapy) Radioisotopes are used in nuclear medicine procedures for imaging and therapy. Imaging isotopes emit gamma rays, therapeutic isotopes emit low-energy gamma rays or high-energy beta or alpha radiation. Radioisotopes are chelated to the targeting monoclonal antibodies or proteins to form radio-immunoconjugates (RIC).

A number of chelating agents are used. These bifunctional molecules have a covalent bond to the antibody and form a cage to hold the radioisotope at the other end of the molecule. The more arms in the cage, the more stable the chelation of the radioisotope. If the radioisotope breaks loose, then it may end up in the kidneys or bone marrow, causing delayed radiation damage to these organs.

A new approach to therapy is emerging where radioisotopes that emit very short range (80 μm) alpha particles with high LET are tagged onto monoclonal antibodies for targeted alpha therapy (TAT) to form alpha-immunoconjugates (AIC).

**Abbreviations:** AIC = Alpha-immunoconjugates, LET = Linear energy transfer, RIC = Radio-immunoconjugates and TAT = Targeted alpha therapy.

**Related Article:** Targeted alpha therapy

**Radioisotope**

(Nuclear Medicine) Isotopes that are unstable and disintegrate (undergo radioactive decay) via emission of a particle and/or γ radiation are called radioisotopes. The nuclear composition of a
Radioisotope cameras

Radioisotope scanner collimator line source response

Radioisotope cameras

(Nuclear Medicine) In the 1950s and 1960s radioisotope scanners were replaced with radioisotope cameras. One of the first ideas was published by the Swedish group from Lund University where Johansson and Skanse reported on the use of a multi channel collimator, solid sodium iodine crystal, and blue-sensitive film to obtain pictures of radioactive distributions in vivo. The sensitivity was very low and not practical for clinical use.

In 1956 Hal O. Anger proposed the first scintillation camera where the read out of the scintillation light from the scintillation crystal was done electronically with PM-tubes and an electronic circuit. The camera was produced commercially and several cameras were produced in the mid 1960s. The first scintillation camera (or Anger camera) in Europe was installed at Lund University Hospital in Sweden.

The first camera was equipped with a pinhole collimator and with only nine PM-tubes.

The modern scintillation cameras have the same basic principles as the Anger camera including the so-called Anger electronic positioning system. Usually these cameras are called gamma cameras.


Related Articles: Anger camera, Scintillation camera, Gamma camera

Radioisotope scanner

(Nuclear Medicine) Radioisotope scanning is also known as rectilinear scanning or scintigraphy. After the introduction of thyroid uptake measurements with radioiodine the scanning technique was developed for the thyroid to identify hot and cold nodules. Thereafter scanners were constructed for scanning larger areas of the body. Then mostly the lungs, liver, kidneys, heart and brain were investigated.

For a scanner a high sensitivity and high spatial resolution are desirable. In general the higher the spatial resolution the lower the sensitivity (mostly depending on the collimator design).

A radioisotope scanner consists of a focusing collimator and a scintillation crystal to convert the photon energy into a signal. In the conventional mechanical scanner a system of motors and controls moves the detection system with respect to the patient and synchronizes the readout with the position of the detector. The scanning speed and the line spacing are variable in the scanner.

The total information contained in a scan is proportional to the scanning time and independent of the actual number of lines so long as they are not spaced further apart than the resolution distance of the collimator.

The readout systems in the earlier models were based on dot records where a stylus was moving over a piece of paper and stamping a dot or a bar for a fixed number of input pulses. The density of dots was then proportional to the activity in the collimators’ field of view.

Another system of readout is photographic recording where a light is moved over a photographic film and focused to a point. The light intensity is proportional to the count rate.

In recent systems the reading is stored in a computer memory and the usual image processing and imaging display tools can be used.


Related Articles: Scanning, Scintigraph, Activity distribution

Radioisotope scanner collimator

(Nuclear Medicine) For a radioisotope scanner the purpose of the collimator is to limit the field of view of the detector. In this way spatial resolution is achieved.

For a collimator the point source response is determined. Most collimators are constructed such that the holes are focused to one point and thus define a focal plane for the scanner.

From point source response curves at different depths, a point source response pattern may be constructed. If the response is symmetrical about the axis, this completely determines the spatial collimator response.


Related Articles: Radioisotope scanner, Collimator

Radioisotope scanner collimator line source response

(Nuclear Medicine) For a radioisotope scanner the performance of the collimator to a line source in a plane perpendicular to the axis of the collimator is important. The collimator scans on the x-axis across

FIGURE R.8 Line of stability.
the line source, yielding a bell-shaped response function as with the point source. This shape is identical with the point source function if we have a Gaussian distribution for the x- and y-dependence.

**Related Articles:** Radioisotope scanner, Scintigraph, Collimator


**Radiological technologist**
*(Diagnostic Radiology)* See Radiographer

**Radiologist**
*(Diagnostic Radiology)* See Diagnostic radiology

**Radiology information system (RIS)**
*(Diagnostic Radiology)* Radiology information systems (RIS) are specialized types of information systems for application in radiology departments. This is a database driven distributed system intended to handle the data entry, storage, exchange and access of radiology related patient data and images. RIS works either as a standalone system or as an integrated part of a hospital information system (HIS). RIS integrates different modules for patient data tracking, image handling and reporting:

- Patient registration data
- Patient tracking records
- Result from examinations
- Scanning/imaging data
- Other patient or examination related documents and reports
- Administrative data

Radiology information systems support data exchange and interfaces to HL7, DICOM, PACS.

**RIS/HIS (Radiology/Hospital Information System)**
*(Diagnostic Radiology)* See Radiology information system (RIS) and Hospital information system (HIS)

**Radiolucent**
*(Diagnostic Radiology)* Radiolucent is the characteristic of an object or material that permits radiation, typically x-radiation, to penetrate or pass through it.

**Radiolysis**
*(Radiation Protection)* Although DNA is the critical target for damage to cells caused by ionising radiation, most physical interactions between the radiation and the cell occurs in the cytoplasm involving the absorption of the energy of the radiation by cellular water molecules. The consequences of these interactions may be described by considering the radiolysis of water:

In the pre-chemical stage, the $\text{H}_2\text{O}^+$ and $\text{H}_2\text{O}^*$ radical ions disassociate to produce hydroxyl ($\text{HO}$) and hydrogen ($\text{H}$ or $\text{H}^*$) ions.

These hydroxyl and hydrogen ions are highly reactive.

In the chemical stage the products of chemical reactions in the pre-chemical stage migrate in pairs close enough to react chemically. This process leads to a series of chemical reactions to produce, for example peroxyde ($\text{H}_2\text{O}_2$) and hydroxide ($\text{OH}$) – highly oxidising chemical reagents – as well as water ($\text{H}_2\text{O}$). If the products do recombine as water then it can be assumed that no further harm is done.

At the biological stage, once all the free radicals produced have reacted, the resultant oxidising agents can migrate to cause direct damage to DNA by reacting and reorganising molecule bonds between oxygen and hydrogen atoms in the DNA molecule. This biological damage to the cell can be effected in hours, but may take days, weeks, months or years.

**Related Articles:** Radiation damage, Radiobiological models

**Radionuclide generators**
*(Nuclear Medicine)* Radionuclide generators are used to produce short-lived radionuclides in centres remote from the site of production, using their longer-lived parent radionuclide. The most common generator in use is the $^{99}\text{Mo}/^{99m}\text{Tc}$ generator, which produces $\text{NaTcO}_4$ which can then be used directly in labelling procedures. A schematic cross-section of the generator can be seen in Figure R.9.

The decay series of $^{99}\text{Mo}$ is as follows:

$$^{99}\text{Mo} \rightarrow ^{99m}\text{Tc} + \beta^- + \nu \quad t_{1/2} = 66 \text{ h}$$

$$^{99m}\text{Tc} \rightarrow ^{99}\text{Tc} + \gamma \quad t_{1/2} = 6.02 \text{ h}$$

$^{99m}\text{Tc}$ is being continuously produced by the decay of $^{99}\text{Mo}$, but continuously destroyed by its own decay. As $t_{1/2} (\text{Mo}) >> t_{1/2} (\text{Tc})$, a transient equilibrium is established where after reaching a maximum value, the activity of the daughter follows that of the parent. At this stage, the ratio of the activities of the parent and daughter becomes constant, and in effect the daughter radioactivity decays with an apparent half-life of the parent radionuclide rather than its own. The generator can be eluted every 24 h and one can still obtain maximum activity. However a reduced yield may be eluted before

**Figure R.9** The $\text{Mo}-99/\text{Tc}-99m$ generator.
Radionuclide imaging


Radionuclide imaging

This page contains a review of radionuclide imaging in nuclear medicine. It discusses the production, purity, and therapy of radionuclides used in nuclear medicine. The text covers the clinical use of radionuclides, including radionuclide uptake in tumour cells.

Radionuclide production

The production of radionuclides used in nuclear medicine is typically done in nuclear reactors or accelerators. Radionuclides are used for imaging and therapy. One of the fundamental properties of the nuclides used in nuclear medicine is their half-life, which determines the time they can be used. Radionuclides with a shorter half-life are used for imaging, while those with a longer half-life are used for therapy.

Radionuclide purity

The purity of a radionuclide is important for its clinical use. Impurities can affect the imaging or therapy outcomes. Purity can be measured using gamma spectroscopy or mass spectrometry. Maintenance of purity is essential to ensure the radionuclide's effectiveness.

Radionuclide uptake in tumour cells

Radionuclides are used in nuclear medicine to target and destroy tumour cells. The ideal radionuclide for tumour therapy must accumulate in the tumour cells and be efficiently delivered. This requires a detailed understanding of the therapeutic properties of the radionuclide.

Further Reading:


Related Articles:

GMP, Quality control, Radionuclide purity, Chemical purity, Biological purity

Further Readings:

Radionuclides

Radionuclides

(Nuclear Medicine) Radionuclide is the term used to describe a nuclide which is in the process of undergoing radioactive decay. Associated with each radionuclide is a set of characteristic properties. These include mode of decay, transition energy and half-life.

Diagnosis: For nuclear medicine diagnosis the aim is to provide clinical information while exposing the patient to minimal radiation. Therefore the desirable properties of radionuclides used in diagnosis are as follows:

- Half-life approximately equal to the duration of the investigation
- No particulate radiation
- Emission of a gamma ray of energy high enough to travel through tissue and low enough for efficient detection by a gamma camera (approx. 100–300 keV)
- Readily available

An example of such a radionuclide is 99mTc.

Therapy: The aim here is to destroy diseased cells with radiation; therefore, it is important to locate the radionuclide precisely within the tissue to be treated. The desirable properties of radionuclides used in therapy are as follows:

- Half-life short enough so that hospital stay is limited
- Emission of particulate radiation with penetration similar to the size of the lesion to be treated
- Emission of a gamma ray in order to visualize the radionuclide concentration in the target tissue

An example is 131I.

Radionuclides in radiotherapy

(Nuclear Medicine) The number of radionuclides that are suitable for radiotherapy are limited to the isotopes with suitable physical and chemical characteristics, availability and production procedures.

The physical properties of any radionuclide refer to the radionuclide half-life and emission spectra. Radionuclides that emit electrons or high-LET radiation are best suited for therapy purposes since the energy is deposited close to the decay point and not in distant organs.

The ideal radionuclide has three favourable chemical characteristics: (1) it should be simple to separate a carrier-free sample without polluting isotopes; (2) easy to label it to a tracer and (3) show in vivo stability.

To minimise the patient dose the half-life of the radionuclide should be low. Therefore, the ideal production site is an in-house accelerator where the activity losses during transport are low compared to longer transport from distant production sites.

Related Article: Carrier-free sample

Radiopacity

(Diagnostic Radiology) Radiopacity is the characteristic of an object or material that impedes or attenuates the passage of radiation, typically x-radiation, through it.

Radiopaque markers

(Diagnostic Radiology) Radiopaque markers are typically made of lead (or lead rubber) and placed somewhere in the x-ray beam so they are visible in the image. Letters of the alphabet, ‘L’ and ‘R’ are used to mark the left and right sides of a patient, especially in mammography. Markers might be used to show the location of specific anatomical features, such as scars on the surface of a patient that could be visible in a radiograph.

Radiopharmaceuticals

(Nuclear Medicine) Radiopharmaceuticals are pharmaceuticals labelled with different radionuclides, for use in the field of nuclear medicine. They contain tracer amounts (from roughly one tenth of a ng to a few hundred μg) of molecules or biological agents. The applicability of a radiopharmaceutical is determined by the characteristics of the pharmaceutical part, that is, the main localization and metabolism in a given organ or tissue and the emission properties of the radionuclide. Nearly 95% of the radiopharmaceuticals are used for non-invasive imaging using photon-emitting radionuclides, whereas the rest are for therapy using beta-emitting radionuclides.

The most used radiopharmaceuticals for imaging are based on the use of the radionuclide 99mTc. Prefabricated kits containing the nonradioactive chemicals are used to produce a specific 99mTc-radiopharmaceutical after adding the required activity of 99mTc-pertechnetate. Some other radionuclides, such as 111In, may be labelled to chelating agents. These are claw-like molecules that bind with a metallic atom in the middle of its structure and label a particular molecule, for example an antibody or a blood cell.

Since radiopharmaceuticals are administered to humans, they have to be sterile and pyrogen free. For human use of a radiopharmaceutical it must be approved by the regulatory authority of the country in which it is used. The most important considerations are manufacturing practice, radiation safety issues and aseptic dispensing conditions. Standardized specifications that define the quality of about 60 radiopharmaceuticals for diagnostic and therapeutic purposes are stated in the European Pharmacopeia.

Besides the term ‘radiopharmaceutical’, which is the most common in use, other terms like tracer, radiotracer, and radiodiagnostic/therapeutic agent are used in the literature. It should be noted that the term ‘radiochemical’ should not be mistaken for a radiopharmaceutical, since the former is not sterile and pyrogen free and not usable for human administration.

Related Articles: Kits, Nuclear medicine


Radiopharmacy

(Nuclear Medicine) The radiopharmacy is a specialist pharmacy service involved in the production of radiopharmaceuticals for nuclear medicine.

Most of the radiopharmaceuticals prepared in a radiopharmacy are based on the use of Tc-99m. The eluate from the technetium generator is added to cold kits according to manufacturers’ instructions.

The design of a radiopharmacy falls under the regulatory constraints of the country in which it is operated. The most important considerations are manufacturing practice, radiation safety issues and aseptic dispensing conditions.

Related Article: Radiouclide generators
**Radiosensitivity**

(Radiotherapy) Radiosensitivity describes the relative sensitivity of different cells or tissues in the body to damage caused by ionizing radiation. This relative sensitivity is expressed by using tissue weighting factors to convert the equivalent dose to a tissue or organ to the effective dose to the whole body in order to then estimate the risk to the person of developing a stochastic effect (i.e. cancer or hereditary disease).

A cell's susceptibility to radiation varies with the phase of the cell cycle: least sensitive to radiation late in the S phase and most sensitive in the M and G2 phases. For cells with long cell-cycle times and significantly long G2 phases, there is a second peak of radio-resistance early in G1.

The work of two French radiobiologists Bergonie and Tribondeau, published in 1906, indicated that the more rapidly a cell is dividing, the more sensitive it is to radiation. This is often called the law of Bergonie and Tribondeau and applies well to rapidly dividing cells such as haematopoietic stem cells, which are radiosensitive. Indeed, it has been shown that the most sensitive cells are those that are undifferentiated, well nourished, divide quickly and are highly metabolically active. Amongst the mammalian cells, the most sensitive are spermatogonia and erythroblasts, epidermal stem cells and gastrointestinal stem cells. The least sensitive are neurons and muscle fibres, that is differentiated cells that do not proliferate. However, the law does not apply in all cases. For example, oocytes and lymphocytes are non-proliferating and yet they are very radiosensitive.

A cell's radiosensitivity can be determined from cell survival curves. These describe the relationship between the radiation dose and the proportion of cells that survive following irradiation. The shape of survival curves is commonly described by the linear-quadratic model with the ratio of its two parameters, α and β, often used as a measure of a cell's radiosensitivity. High α/β-values represent tissues with limited potential for recovery from radiation damage, for example skin and most tumours, whereas normal tissues with low α/β-values are characterised by their high recovery potential, for example kidney and spinal cord.

Other factors affect how sensitive a cell is to radiation. Based on the law of Bergonie and Tribondeau, rapidly dividing tumour cells may be expected to be more sensitive to radiation than normal tissue cells. However, this is not always true: tumour cells can be hypoxic and therefore less sensitive to radiation that indirectly damages DNA through free radicals produced by the ionisation of oxygen. Therefore, the presence or absence of oxygen dramatically influences the biological effect of sparsely ionising radiations such as x-rays but there is a much-reduced effect for densely ionising radiations such as α-particles. For more information on the effect of oxygen see the article on Oxygen enhancement ratio.

The lethal effects of radiation can be increased by the administration of radiosensitising agents. Many chemical and pharmacological agents that modify the response of mammalian cells have been discovered but most offer no practical gain in radiotherapy since they affect tumours and normal tissues alike. There is no point in employing a drug that increases the sensitivity of tumour cells if it also increases the sensitivity of normal cells to the same extent. Two types of sensitisers have demonstrated a differential effect between tumour and normal cells and have found practical use in clinical radiotherapy. These are *halogenated pyrimidines* (differential effect is based on the premise that tumour cells cycle faster than normal cells and therefore incorporate more of the drug than the surrounding normal tissues) and *hypoxic-cell sensitisers* (differential effect based on the premise that hypoxic cells only occur in tumours and not normal tissues). A number of clinical trials are currently in progress using the agents Gemcitabine and Temozolomide.

Conversely, the differential effect of radiation response between tumour and normal cells may be increased by the administration of radioprotecting substances (radioprotectors). These agents protect normal cells from radiation-induced damage by promoting cell repair. Currently, the only well established drug is Amifostine which reduces the radiation associated dry mouth effect (xerostomia) observed in patients receiving radiotherapy for head and neck tumours.

**Abbreviations:** AT = Ataxia telangiectasia and DSB = Double-strand DNA breaks.

**Related Articles:** Alpha beta ratio, Cell cycle, Oxygen enhancement ratio, Repair, Repopulation, Redistribution, Reoxygenation, 5Rs of radiobiology, Surviving fraction


**Radiosensitisers**

(Radiotherapy) Radiosensitisers are agents that increase the radiosensitivity of cells and therefore increase the lethal effects of radiation if administered in combination with the radiation.

Over the years, many chemical and pharmacological agents have been discovered that modify the response of mammalian cells to radiation. The simplest of these, and the one that possibly produces the most dramatic effect, is oxygen. It is known that tumour cells can be hypoxic and therefore less sensitive to radiation that indirectly damages DNA through free radicals produced by the ionisation of oxygen. Therefore, the presence or absence of oxygen drastically influences the biological effect of sparsely ionising radiations such as x-rays but there is a much-reduced effect for densely ionising radiations such as α-particles. For more information on the effect of oxygen see the article on Oxygen enhancement ratio.

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**Related Articles:** Oxygen enhancement ratio, Reoxygenation, Repair

Radiotherapy

(Radiotherapy) Radiotherapy is the treatment of a disease with ionizing radiation. The term ‘radiation therapy’ is also used.

Depending on the distance between the radiation source and the target volume, that is the tissues to be treated, radiotherapy is divided into two categories: teletherapy and brachytherapy.

In teletherapy the source is far from the target (Greek word tele, far away), while the source is placed close to the target in brachytherapy (Greek word brachy, short).

In teletherapy historically x-ray tubes operated at higher voltages have been used (orthovoltage therapy). Now electron linear therapy (Greek word brachy, short).

Far away), while the source is placed close to the target in brachytherapy (Greek word brachy, short).

Conventional brachytherapy uses sealed sources, and conventional brachytherapy is the subject covered under this heading.

(Radiotherapy, Brachytherapy)

Related Articles: Brachytherapy, Teletherapy

Radiotherapy

(Magnetic Resonance) In MRI, pulsed radiofrequency (RF) energy is deposited into the investigated sample for spin excitation. In order to create the desired spin excitation, the frequency is calculated using the Larmor relation which relates the so-called resonance frequency to the main magnetic field. For protons, this resonance frequency is 42.6 MHz/T, placing the RF pulses in the same frequency range as of the electromagnetic fields used for generating and detecting radio waves.

See also Radiofrequency.

Related Articles: Radiofrequency, Larmor frequency

Radium

(Radiotherapy, Brachytherapy) Radium-226 is a naturally occurring radionuclide belonging to the decay series uranium-238 – lead-206. Radium decays (alpha) with a half-life of 1600 years into radon-222, an inert gas with a half-life of 3.82 days. The decay of radium-226 via radon-222 to stable lead-206 results in a number of photons with energies up to 2.45 MeV, beta particles (maximum energy 3.3 MeV) and alpha particles. A radium source (a radium salt mixed with a filler) with a platinum encapsulation, that is filtered by 0.5 mm platinum, has an average photon energy of 0.83 MeV, when the radium is in equilibrium with its daughter products (the exposure weighted average energy is 1.25 MeV).

The beta and alpha particles from the decay are absorbed in the encapsulation.

Types of Radium-226 Sources: Radium was initially used for temporary low dose rate implants, its specific source strength is low, and even low dose rate sources were large in size. Radium sources were available as needles and tubes with different dimensions and source strengths.

Needles and tubes were often loaded into other types of applicators. Figure R.10a shows an applicator for treatment of cervical cancer, consisting of an intrauterine probe and a box. The box was loaded with a number of needles, see Figure R.10b. (This specific coupled applicator had source strengths of 3.3 GBq + 4.1 GBq [90mCi + 110mCi] apparent activity for the probe and the box respectively. The dose rate at point A was 2.0 Gy/h. Dimensions: probe length 6 cm and diameter 6.5 mm, box size 4.5 × 4.5 × 0.6 cm3.)

A radium source, if damaged, could leak the toxic radium salt with its decay products, including the radioactive radon gas. The high photon energy further constitutes a radiation protection problem. Radium sources are in principle not used today, but the vast amount of clinical experience gained with brachytherapy using radium sources still makes an impact on modern brachytherapy practises.

Radon-222 Sources: The radioactive gas radon-222 was the first isotope used for permanent implants. The gas from the decay of radium-226 was trapped and encapsulated as a small seed. The radon source has a very short half-life, 3.82 days, and the same photon spectrum as radium-226.

Related Articles: Brachytherapy sources, Radium substitute isotope, Intracavitary brachytherapy

Radium substitute isotope

(Radiotherapy, Brachytherapy) In the ICRU Report 38 (1985), Dose and VolumeSpecification for Reporting Intracavitary Therapy in Gynaecology, the following is stated:

‘The replacement of radium by 137Cs, 192Ir and 60Co may be accomplished according to two options. In the first option, the new sources (mainly 137Cs) are similar in size and have an output similar to radium sources. The same technique of application can then be applied and the clinical experience gained with radium remains fully relevant. The principal advantages of the replacement of the radium concerns radiation protection; these include no contamination from leakage and less shielding in the case of 137Cs and 192Ir’.

Compare also the source strength specification ‘mg Ra eq’.

FIGURE R.10  (a) Lund applicator for cervical cancer treatments, intrauterine probe and box, coupled for secure positioning with the probe perpendicular to the box, and (b) a radium applicator box showing the loading pattern for radium needles, one dummy radium needle loaded.
Radon (General)

<table>
<thead>
<tr>
<th>Property</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Symbol</td>
<td>Rn</td>
</tr>
<tr>
<td>Element category</td>
<td>Noble gas</td>
</tr>
<tr>
<td>Mass number A</td>
<td>222</td>
</tr>
<tr>
<td>Atomic number Z</td>
<td>80</td>
</tr>
<tr>
<td>Atomic weight</td>
<td>222 kg/kg-atom</td>
</tr>
<tr>
<td>Electronic configuration</td>
<td>1s^2 2s^2 2p^6 3s^2 3p^6 3d^{10} 4s^2 4p^6 4d^{10} 4f^{14} 5s^2 5p^6 5d^{10} 6s^2 6p^6</td>
</tr>
<tr>
<td>Melting point</td>
<td>202 K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>211.3 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>9.73 kg/m^3</td>
</tr>
</tbody>
</table>

**History:** Radon was discovered in 1900 by Friedrich Dorn as one of the products of the radioactive decay of radium. At room temperatures it is a dense colourless odourless gas which is radioactive, undergoing alpha decay. The least unstable isotope of radon, radon-222, has a half-life of about 3.8 days. Radon is produced naturally by successive decays of thorium and uranium found in some igneous rocks. The radon gas seeps to the surface where, due to its high density, it can become trapped in buildings and can, in certain locations, accumulate to levels that represent a significant radiological health hazard.

**Medical Applications:** In early years, radon (222Rn) seeds prepared by capturing the radon gas from radium (a radon generator) were used in cancer therapy. Radon seeds were subsequently replaced by other sources such as 198Au and 121I.

**Related Articles:** Radioactivity, Radioactive materials, Radioactive decay, Alpha emission

**RAID technology**

(Diagnostic Radiology) RAID is an abbreviation that stands for redundant array of independent disks. The RAID technology refers to different hardware and/or software methods of data storage which allow increased efficiency, reliability and performance by simultaneous usage of two or more hard disk drives (RAID array).

Each RAID array combines the different hard drives into one logical drive, thus achieving one of the following – mirroring, striping, error correction.

<table>
<thead>
<tr>
<th>RAID</th>
<th>Configuration</th>
<th>RAID Data Distribution</th>
<th>RAID Performance</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>Stripping</td>
<td>RAID 0 combines two or more RAID 0 arrays (strip) into RAID 1 array (mirror), thus increasing capacity, speed and reliability.</td>
<td>RAID 0: Increased capacity, speed and reliability.</td>
</tr>
<tr>
<td>1</td>
<td>Mirroring</td>
<td>RAID 1 uses two or more identical disks drives, each one storing the same copy of data. The hard drives in RAID 1 are independent and replaceable in case of future.</td>
<td>RAID 1: Increased reliability.</td>
</tr>
<tr>
<td>5</td>
<td>Stripping</td>
<td>RAID 5 combines two or more RAID 0 arrays (strip) into RAID 5 array (mirror), thus increasing capacity, speed and reliability.</td>
<td>RAID 5: Increased reliability.</td>
</tr>
<tr>
<td>01</td>
<td>Stripping and Mirroring</td>
<td>RAID 01 combines two or more RAID 0 arrays (strip) into RAID 1 array (mirror), thus increasing capacity, speed and reliability.</td>
<td>RAID 01: Increased capacity, speed and reliability.</td>
</tr>
</tbody>
</table>

**RAM memory**

(General) The RAM (random access memory) of one computer system is a memory allowing direct access to any byte on the chip. Usually the RAM is a volatile type memory, which requires constant refreshing (e.g. every several ms), but has very short access time (e.g. every several ns). Usually all data processed by the CPU is temporarily stored into the RAM, this way the amount of RAM is of prime importance for the speed of data processing. There are many different types of RAM.

**Ramp converter**

(General) A ramp converter is a type of analogue to digital converter (also called linear ramp converter, ramp-compare, integrating, dual-slope, multi-slope or Wilkinson type ADC) used in radiation detection systems. Its purpose is to convert the analogue pulse into a digital signal. It does so by comparing the amplitude of an incoming pulse with that of a pulse generated by a linear ramp. The input pulse is used to charge a capacitor, and discharge time, which is proportional to pulse amplitude (proportional to radiation energy), is measured using a clock oscillator. The number of the clock pulses counted is proportional to the discharging time, which in turn is proportional to the radiation energy.

**Ramp filter**

(Nuclear Medicine) A ramp filter is the basic filter used in the reconstruction technique of filtered back projection. It is used with simple back projection to eliminate the 1/r blurring effect.
Ramp time

(Magnetic Resonance) In MRI, the term ramp time denotes the time interval for a magnetic field gradient, frequently measured in millitesla per meter or mT/m, to build up by driving a current through one or more of the physical gradient coils.

The ramp time during the build-up phase is also frequently denoted rise time. Ramp times of course also occur when a magnetic field gradient is turned off (see Figure R.13).

The ramp time to maximum gradient amplitude of the physical gradients is, together with the maximum amplitude itself, an important characteristic of an MRI system, since this time influences, for example minimum achievable echo times in imaging pulse sequences.

In a conventional pulse sequence, gradients in the slice, phase and readout (frequency) direction are mandatory and each of these gradients are associated with two ramp times (up and down).

In modern MRI scanners, minimum ramp times to maximum gradient strength are on the order of a few hundreds of microseconds.

The parameters minimum ramp time and maximum gradient strength are governed by health aspects, since when a magnetic field variation dBldr is built up or turned off a temporally varying magnetic field dBldr interacts with the body according to

$$\frac{dB}{dt} = \frac{dB}{dr} \frac{L_r/2}{\Delta t} = \text{SR} \cdot \frac{L_r}{2} \tag{R.1}$$

In the previous equation, dBldr denotes the magnetic field variation over time, dBldr is the intended gradient strength in the r direction after ramping up, L_r is the effective length of the gradient coil in the r direction and ∆t is the ramp time for the gradient to reach the value dBldr. SR denotes slew rate and is defined as the ratio between dBldr and ∆t.

Related Articles: Gradient coils

RANDO phantom
(Radiation Protection) RANDO is a popular body phantom (anthropomorphic phantom), trade mark of The Phantom Laboratory. The phantom is used for assessing dose distribution, mainly in radiotherapy and diagnostic radiology. The phantom is constructed of tissue-simulating material (with natural human skeleton) and mimics human body. It is sliced at 2.5 cm sections and each section includes openings for inclusion of dosimeters (e.g. TLD tablets). There are two main types of phantoms – RANDO man and RANDO woman.

Related Article: Anthropomorphic phantom
Further Reading: http://www.phantomlab.com/rando.html

Random coincidence
(Nuclear Medicine) A coincidence between two photons originating from different annihilation positions.

In PET imaging the coincidences are localized along a line of response (LOR). The LOR is drawn between two detectors with a near-simultaneous detection. This is possible because of the use of positron emitting nuclides. When a positron is brought to a near halt, it annihilates together with an electron, emitting two photons in the process. The annihilation photons are emitted back to back, that is ~180°. Following event detection in a detector, a number of opposite detectors are scanned in order to detect the corresponding annihilation photon, that is there is a projection for each detector.

If both annihilation photons travel through the object without interacting with the surrounding materials before being detected, the coincidence is referred to as a true coincidence.

A second coincidence type is random coincidences. They occur when photons from two different annihilations are registered in opposite detectors near-simultaneously. The occurrence of random coincidences increases with the square of the administered activity, unlike true coincidences that increases linearly.

More information about event types in PET can be found in the article Event type in PET.

Related Articles: PET, Event type PET, True coincidences, Scatter coincidences

Random noise
(Nuclear Medicine) See White noise

Range energy relationship
(Radiotherapy) See Electron practical range

Rapid acquisition relaxation enhancement (RARE)
(Magnetic Resonance) Rapid acquisition relaxation enhancement (RARE) is a pulse sequence characterized by a 90° pulse followed by a series of rapidly applied 180° rephasing pulses and multiple echoes. See Fast spin echo (FSE) for a more detailed description.

Related Articles: Echo train length, Fast spin echo (FSE), Half-acquisition single-shot turbo spin echo (HASTE), Turbo spin echo (TSE)
RARE (rare acquisition relaxation enhancement)
(Magnetic Resonance) See Rapid acquisition relaxation enhancement

Rare earth metals
(Diagnostic Radiology) The rare earth metals are a series of about 17 elements in the periodic table (lanthanides series with atomic number from 58 to 70) that include lanthanum, gadolinium, and ytterbium. The significance in radiology is that compounds of several of these elements are fluorescent materials used in intensifying screens.

See Rare earth screen for more details.

Rare earth screen
(Diagnostic Radiology) Up until the 1970s calcium tungstate (CaWO₄) was the typical fluorescent material in radiographic intensifying screens. These have been replaced by a variety of materials/compounds with a rare earth element (such as lanthanum, gadolinium, and ytterbium) as the x-ray absorber. A major advantage is a higher x-ray absorption efficiency than tungsten, because the lower atomic numbers (Z), places the K edge at a more effective energy with respect to the typical x-ray spectrum.

Some more typical rare earth phosphors include lanthanum oxybromide and gadolinium oxysulphide. Depending on the activator added to the rare earth screen the colour of the emitted light may change. For example, activators with thulium emit blue light, activators with terbium emit green light, etc. yttrium oxysulphide is suitable for blue-sensitive films as it emits blue light.

Rayleigh distribution
(Magnetic Resonance) The Rayleigh distribution, or more precisely, the Rayleigh probability distribution function, is given by

\[ \text{PDF}(B) = \frac{B}{\sigma^2} e^{-B^2/(2\sigma^2)} \]

where \( B \) and \( \sigma \) are parameters. This distribution is encountered when the amplitude from a large number of scattering objects is studied, see Figure R.16. In a pulse echo system, each scatterer in a resolution cell will contribute with a signal that can be described by its phase and amplitude in a complex phasor diagram, as in Figure R.17. The sum of all contributions \( B_j \) sum up at the receiver to \( B_{tot} \). If all phases are uniformly distributed over \([-\pi, \pi]\), and the number of scatterers is large, it can be shown that the real and imaginary parts are Gaussian random variables with the same variances, and zero means. Through a transformation it can further be shown that the phasor magnitude \( B_{tot} \) is Rayleigh distributed according to the previous relation, where

\[ \sigma^2 = \lim_{N \to \infty} \frac{1}{2N} \sum_{j=1}^{N} |B_j|^2 \]
Rayleigh scattering

(Radiation Protection) Rayleigh scattering (named after Lord Rayleigh, 1842–1919) is the elastic scattering of light or other electromagnetic radiation by particles much smaller than the wavelength of the light. In this process the incident photon interacts with an absorber atom as a whole and this implies that all orbital electrons of the absorber atom are tightly bound and contribute coherently to the scattering process. The energy of the incident photon is conserved, but its direction of propagation is changed.

Related Article: Elastic scattering

Rayleigh scattering

(Ultrasound) Rayleigh scattering is said to occur when sound (or light) is scattered by particles much smaller than a wavelength. ‘Much smaller than’ in practice usually means about one tenth of the wavelength. Typical for Rayleigh scattering is that the intensity varies as the fourth power of frequency (up to the point where the scatterer becomes large compared to the wavelength, and it is no longer Rayleigh scattering). The name comes from Lord Rayleigh, who showed that the sky appears blue due to sunlight being scattered in the air, blue representing the shortest wavelengths, or equivalently, the highest frequencies. Scattering in this regime has important implications in ultrasound imaging. Tissue is often modelled as subwavelength scatterers, as well as blood cells, used in modelling of Doppler signals.

RBE (relative biological effectiveness)

(Radiotherapy) See Relative biological effectiveness (RBE)

RC circuit

(General) An RC circuit (resistor–capacitor circuit or RC filter) is an electric circuit composed of resistor(s) and capacitor(s) driven by a voltage or current source. The simplest example of an RC circuit is a circuit composed of one resistor and one capacitor in series. The charged capacitor will discharge its energy into the resistor. This voltage across the capacitor over time could be found through Kirchhoff’s current law, by which the current coming out of the capacitor must equal the current going through the resistor. This results in the linear differential equation (left). When solved, it results in the exponential decay function (right):

\[ CV \frac{dV}{dt} + \frac{V}{R} = 0 \]

When solved, it results in the exponential decay function:

\[ V(t) = V_0 e^{-t/RC} \]

where

- \( V_0 \) is the input voltage, and the product of the circuit resistance \( R \) (in ohms)
- The circuit capacitance \( C \) is the time constant \( \tau \) (in seconds)

RC circuits can operate as filters (high-pass or low-pass), integrators or differentiators (Figure R.18).

At high frequency, that is when \( \omega > 1/RC \), the RC circuit operates as integrator. The capacitor has insufficient time to charge up and so its voltage is very small. Thus the input voltage approximately equals the voltage across the resistor.

At low frequency, that is when \( \omega < 1/RC \), the RC circuit operates as differentiator. The capacitor has time to charge up until its voltage is almost equal to the source’s voltage.

Related Articles: Circuit, Circuit(s), Electrical, Delay circuit

RC time constant

(General) See RC circuit

Reaction time

(General) Reaction time is the speed at which we are able to process information and make decisions. Reaction time is the time between the onset of the stimulus and the start of the movement in response to it. Mean reaction time for young adults is approximately 190 ms to detect a visual stimulus, and approximately 160 ms to detect an auditory stimulus.

Rayleigh scattering

An illustration of a random walk realization, where a number of complex phasors \( B_j \) with different phase and amplitude add up to the phasor \( B_{tot} \).

A histogram of occurrences of amplitude values for the received signal from a large number of point sources. The grey line is a fitted Rayleigh distribution.

Rayleigh scattering

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In technology, response time is the time a system or functional unit takes to react to a given input.

**Further Reading:** Kosinski, R. J. 2008. A literature review on reaction time, Clemson University. http://biae.clemson.edu/bpc/Lab/110/reaction.htm

### Readout gradient

*(Magnetic Resonance)* ‘Readout gradient’ is an alternative term used to refer to the frequency encoding gradient in a conventional MRI pulse sequence.

**Related Articles:** Frequency encoding, Readout period

### Readout period

*(Magnetic Resonance)* This term is used to refer to the period of time when the MRI scanner is acquiring data in the presence of the readout or frequency encoding gradient.

**Related Articles:** Frequency encoding, Readout gradient

### Real-time imaging

*(Magnetic Resonance)* Magnetic resonance imaging (magnetic resonance) is normally regarded as a relatively slow imaging modality, because of the need to acquire multiple signals in order to provide adequate k-space data set for image reconstruction. However, techniques have been available for some time to accelerate the imaging process, the motivation generally being to improve patient acceptability and minimise the likelihood of motion artefacts. Pushing these techniques to their extremes, often combining several different approaches, it is possible to collect images at rates of multiple frames per second (fps), allowing dynamic imaging of physiology (e.g. contrast agent uptake) or kinematic processes (e.g. cardiac motion), or the guidance of interventional procedures.

MRI is always a trade-off between speed (temporal resolution), spatial resolution and signal-to-noise ratio (SNR). The compromises involved in real-time imaging are often detrimental to image quality, introducing a variety of artefacts. The applications of dynamic imaging often do not require the same image quality as would be needed for static diagnostic imaging, but nevertheless there are application-specific image quality requirements that limit the extent to which a particular approach to acceleration can be applied.

The main approaches to acceleration of data acquisition encountered in real-time imaging are as follows:

1. **Multiple k-space lines per shot:** Acquisition of multiple echoes with different phase encoding following a single excitation pulse is a well-established technique, exploited in sequences such as turbo spin echo/fast spin echo and a variety of fast gradient echo methods. In the limit (e.g. single shot echo planar imaging [EPI]), it is possible to traverse the whole of k-space following a single excitation, but image quality can be seriously compromised by poor SNR and marked geometrical distortion.

2. **Reduced coverage of k-space:** Data from the edges of k-space in the phase encoding direction can be omitted, reducing the number of k-space lines that need to be acquired but also compromising spatial resolution. In half-Fourier imaging, data from just under half of k-space are omitted and the missing data points are reconstructed by exploiting symmetries in k-space. This approach compromises SNR.

3. **Partially parallel imaging:** Techniques such as SENSE, SMASH and GRAPPA allow reduction in data acquisition, in real space or in k-space, by exploiting the redundancies inherent in data acquired using RF coils composed of multiple elements. Coils with up to 32 elements are now available commercially, and the feasibility of larger numbers of elements has been demonstrated. As well as coil technology, partially parallel imaging is limited by SNR and artefact issues.

4. **Non-Cartesian k-space trajectories:** Coverage of k-space in, for example radial or spiral trajectories can accelerate imaging or reduce the impact on image quality of other approaches that reduce k-space coverage. However, these techniques introduce artefacts and reconstruction is problematic.

5. **View sharing:** In this approach, also known as sliding window imaging, data are acquired continuously and images are reconstructed more frequently than the whole of k-space is refreshed, so some data are retained over several sequential images.

6. **Temporal redundancy:** This is a related class of technique to view sharing. In keyhole imaging, the centre of k-space is updated in each frame of a dynamic image set, but data from the edges of k-space from the first frame are retained throughout the series. This approach is suitable in situations where image contrast changes over time, but structural information remains static (e.g. contrast agent uptake studies). There are more advanced approaches that update different parts of k-space at different intervals of time in an effort to improve handling of high-resolution data. Emerging techniques such as kt-BLAST take a more sophisticated approach to exploitation of spatiotemporal correlation and redundancy in a way that is mathematically optimised to a particular application through the use of training data.

Real-time imaging is finding particular applications in cardiac MRI, making multiphase imaging over a large volume possible within a single breath-hold (Figure R.19).

**Related Article:** Interventional MRI

### Real-time portal imaging

*(Radiotherapy)* With the availability of digital, electronic portal imaging systems, it is possible to acquire images at rates of several frames per second. These images may be used to obtain movie loop
sequences during the course of radiotherapy beam delivery. These data provide two types of dynamic information:

1. Patient movement during beam delivery – so-called intra-fraction movement
2. Verification of dynamic treatment such as dynamic intensity-modulated radiotherapy (IMRT)

**Intra-Fraction Patient Movement:** This is the motion of anatomy whilst the treatment is being delivered. ICRU recommend the use of an internal margin (IM) to define the internal target volume (ITV) to account for this. Intra-fraction is particularly important for treatment sites where the effects of breathing are significant, such as the lung or liver. Real-time portal imaging provides little soft tissue information and thus implanted fiducial markers are often used to enable determination of the level of intra-fraction motion.

**Verification of Dynamic Treatment:** Examples of this include dynamic IMRT delivery, in which a sequence of images of the changing field shape is acquired and the field shape and intensity in each image is compared with its prescribed values. The distribution of the summed delivery is also compared with prescription. Figure R.20 illustrates an image of an IMRT beam acquired summing a set of real-time portal imaging frames.

**EXAMPLE**

A novel approach to real-time portal imaging has been developed by Shirato et al. They have developed a method of tumour tracking in real time using kV energy portal imaging with two angled kV x-ray sets and x-ray image intensifiers to image fiducial markers at 25 frames per second (see Further Readings).

**Abbreviations:** ICRU = International Commission on Radiation Units, IM = Internal margin, IMRT = Intensity-modulated radiotherapy and ITV = Internal target volume.

**Related Articles:** Portal imaging, Electronic portal imaging, Electronic portal imaging device


**Receiver**

(Ultrasound) The term ‘receiver’ is used to denote the detecting element of an ultrasound transducer. For a pulse-echo system, the same elements are used for transmit and receive; for continuous wave (CW) Doppler, separate transducer elements are required.

**Related Articles:** Pulse echo, CW Doppler

**Receiver coil**

(Magnetic Resonance) The receiver coil in MRI detects RF signal from the patient. The origin of the signal is the transverse component of nuclear spin precession throughout the volume to which the coil is sensitive. The coil detects this signal and feeds it to the system electronics, typically for separation into in-phase and quadrature components, digitization and ultimately for transcription to k-space.

Physical realisations of RF coils range from simple single loop coils to multi-element phased arrays. Receiver coils form part of a tuned circuit designed to resonate at the RF frequency of interest. The copper strips typically used to construct the coil together with added capacitance form the inductive/capacitive elements in the resonant circuit.

In practical application, receive coils are associated with a particular anatomical region of interest, for example head coil, spine coil, spine coil, breast coil, knee coil, endorectal coil, etc. Generally these coils can be connected and disconnected as required by the system operator for a given clinical examination. In addition an MRI system will have a permanently built in coil in the system housing called the ‘body coil’ which acts both as transmit and receive coils. Coils implementing both transmit and receive functions are called ‘transceiver’ coils.

A ‘volume coil’ receiver coil is designed to provide good uniformity of sensitivity to signal throughout the volume imaged, which translates as good image uniformity. Volume coils are used where a relatively large anatomical volume is imaged, for example a head coil or body coil. With a ‘surface’ coil, sensitivity to signal falls off rapidly with distance from the coil. While this reduced field of view gives a poorer uniformity than a volume coil, SNR values are higher over the anatomy of interest. Surface coil designs may be preferred where the anatomy imaged is superficial and can be placed close to the receiver coil (e.g. a spine coil). Phased array coil designs combine many surface coil elements together in order to increase potential extent of anatomy imaged while preserving desirable features of surface coil performance.

**Receiver operating characteristic (ROC)**

(Diagnostic Radiology) The receiver operating characteristic (ROC) method is used to analyse data (images). It takes into account the skills (or the bias) of the observer (the operator). The method is also related with the SNR of the image. The assessment of the images uses the criterion of finding (seeing) an abnormality in the image (e.g. pathological structure in the lungs), when this abnormality actually exists. The possible results in such an observation can be as follows:

- Seeing an abnormality when it exists on the image (true positive, TP)
- Seeing an abnormality, but in fact it does not exist on the image (false positive, FP)

**FIGURE R.20** Summed real-time portal image of an IMRT beam.
• Not seeing an abnormality (all looks normal) when it exists on the image (false negative, FN)
• Not seeing an abnormality and in fact there is no abnormality on the image (true negative, TN)

When a number of images are seen by various observers and their findings are analysed, one can extract two important measures:

• Sensitivity (the ability to detect an abnormal finding in the image) – this is the fraction of abnormal findings, that are actually classified as abnormal. This can be expressed by the formula:

  \[
  \text{Sensitivity} = \frac{TP}{TP + FN}
  \]

• Specificity (the ability to detect a normal finding in the image) – this is the fraction of normal findings, that are actually classified as normal. Sometimes specificity is also called selectivity. This can be expressed by the formula:

  \[
  \text{Specificity} = \frac{TN}{TN + FP}
  \]

The overall accuracy of the exercise can be expressed by the formula

\[
\text{Accuracy} = \frac{TP + TN}{TP + TN + FP + FN + TN}
\]

Plotting the sensitivity as a function of the specificity (for the whole exercise) presents a curve useful for assessing image quality. Usually on the y-axis is the sensitivity (also referred to as true positive fraction, TPF). On the x-axis is \((1 − \text{specificity})\), also referred to as false positive fraction, FPF. The ROC curve received this way is a good quality indicator and can be used to compare various imaging methods (or modalities) for the detection of some pathological findings. The main areas of the ROC curve are shown on Figure R.21.

If this assessment is made by medical physicists they can use a test object (phantom) instead of a real medical image with pathology. The test object used can be one used for assessment of image contrast resolution (or noise). During the exercise the observers mark the visibility of the low contrast inserts of this test object.

Figure R.22 presents three typical ROC curves – line A shows a case when abnormality can not be distinguished from normality. In this case the SNR of the image is almost zero and the observer is guessing the results. Curve B shows improvement as the ability to detect abnormality (or to see the inserts of the test object) increases. The best case is when the curve is close to the axes (as curve C).

The area under the curve is a measure of detectability. In the worst case (line A) the detectability is 0.5. In the perfect case the area under the curve (the detectability) will be equal to 1 (the curve coincides with the axes of the diagram – i.e. above C on Figure R.22).


**Receptor targeting**

* (Nuclear Medicine) Receptor targeting refers to use of radiopharmaceuticals which targets specific cell receptors. Tumour cells with a high density of a specific receptor can be targeted by the bioengineered radiopharmaceuticals. The radiopharmaceuticals will ideally accumulate selectively in malignant cells; hence giving no unnecessary radiation dose to the surrounding tissue.

One example of a molecular imaging agent is 111In-Octreotide. It is an 8-peptide residue that binds to somatostatin subtype-2 peptides. The somatostatin is over-expressed in neuro-endocrine tumours, which makes it an attractive choice for both therapy and imaging. Another example is 99mTc-Depreotide which binds to somatostatin subtypes 2, 3 and 5 with high affinity and these receptors are over-expressed in small cellular lung cancer.

**Related Articles:** Tracers, Analog tracers, Distribution volume, Partition coefficient, Tracer flux between compartments, Tracer kinetic modelling

**Recoil electron**

* (General) A recoil electron is the electron that is ejected from an atom as a result of a Compton interaction with an incident photon. Only some of the energy of the photon is given to the electron, and the photon is scattered.

**Related Articles:** Compton effect, Compton scattering
Recombination correction

Ion recombination in gas-filled radiation detectors, that is ionisation chambers, occurs during collisions between positive ions and free electrons or negative ions resulting in neutral atoms (molecules) being formed. The degree of recombination depends on the geometry of the chamber, the polarising voltage and the rate of ionisation produced by the radiation. Therefore the recombination correction is needed to exposure measurements (dosimetry) made with ionization chambers.

The recombination correction factor $k_{sat}$ used in dosimetric protocols is defined as the inverse of the collection efficiency $f$:

$$k_{sat} = \left(\frac{1}{f}\right)$$

The recombination correction factor is generally estimated from a two-voltage analysis (TVA) for each beam quality (x-ray and high-energy photon beams, electron beams and proton beams). The two-voltage technique proposed by P.R. Almond consists in two measurements of current or charge:

1. $I_1 (Q_1)$ current (charge) for the normal operating bias voltage $V_1$
2. $I_2 (Q_2)$ current (charge) or the voltage $V_2 < V_1$

The ion recombination correction factor $k_{sat}$ is then equal to

$$k_{sat} = \left[\frac{(V_1/V_2)^2 - 1}{(V_1/V_2)^2 - (I_1/I_2)}\right]$$

Recombination correction factors are measured at different dose rates and different polarising voltages during the calibration of chambers in diagnostic radiology and in radiotherapy dosimetry.

**Related Articles:** Exposure, Ionization recombination loss, Ionization chamber


Recombination effect

The charged particles created by ionizing radiation in gas-filled detectors may collide with neutral atoms that are in thermal random motion. There are many types of collisions, including charge transfer collisions. In the charge transfer collisions a positive ion can take an electron from the neutral molecule or a free electron can be attached to a neutral atom (molecule). During the collision between the positive ion and the free electron or a negative ion a neutral atom (molecule) may be formed. This process is called ion recombination.

See Related Articles for further details.

**Related Articles:** Ion recombination, Ionisation recombination loss, Recombination correction, Thimble chamber

Recombination factor

See Recombination correction

Recovery coefficient (RC) in emission CT

Due to the partial volume effect, small objects near the resolution limits of the imaging system appear to have a lower concentration of radioactivity than is actually present. The recovery coefficient (RC) is defined as the ratio of the apparent concentration to true concentration. The RC for a three-dimensional object is the product of the RCs in each dimension.

Theoretically, if the size of a given small object and the resolution of the emission tomography system are known, a recovery coefficient correction factor can be used to correct for the partial volume effect and calculate the actual radioactivity concentration in the object.

**Related Article:** Partial volume effect

Rectangular FOV

Use of a rectangular field of view (FOV) is an approach to reducing MRI image acquisition time in situations in which the body part being imaged is longer in one direction than the other within the desired image plane.

An example is coronal imaging of the head. In such a situation the FOV need not be square: fewer pixels can be acquired in the left–right direction than in the cranio–caudal direction, without the risk of image wrap-around due to aliasing.

This can be achieved without loss of spatial resolution by collecting lines along the phase encoding axis in $k$-space more sparsely. The interval between lines in $k$-space is inversely proportional to the FOV size:

$$FOV = \frac{1}{Dk}$$

Reduction of the number of $k$-space lines equates to reduction in the number of repetitions of the sequence needed for phase encoding purposes, and hence a reduction in the overall image acquisition time. As long as acquisition extends as far out in $k$-space, pixel size and hence spatial resolution will be unaffected.

Because less data are collected, use of a rectangular FOV leads to a reduction in signal to noise (SNR) in the image, according to the square root of the rectangular FOV factor. Thus if the field of view is reduced to 0.75 of its original size along one axis, the resulting SNR will be $\sqrt{0.75} = 87\%$ of its original value.

**Related Articles:** $k$-space, Phase encoding

Rectangular pulse

In magnetic resonance imaging, the rectangular pulses or hard pulses are achieved by applying a time independent $B_1$ field, resulting in a RECT function–shaped (excitation) pulse in the time domain.

Rectangular pulse can be designed to be very short and therefore can be used to excite a broad frequency range. If the small flip angle approximation is fulfilled, the frequency profile of a rectangular pulse is a sinc (i.e. $\sin(\pi x)/\pi x$).

The flip angle $\theta$ of a rectangular pulse is directly proportional to the amplitude of the applied RF field $B_1$ and the pulse duration $T$:

$$\theta = \gamma B_1 T$$

where $\gamma$ is a constant called the gyromagnetic ratio.
Pulses are used for the initial magnetization preparation or for spectroscopy where it is beneficial to excite a wide range of frequencies.

**Related Articles:** $B_0$ field, Gyromagnetic ratio, Magnetic resonance imaging


**Rectification**  
*(Diagnostic Radiology)* See Rectifier

**Rectification, full-wave**  
*(Diagnostic Radiology)* See Rectifier

**Rectification, half-wave**  
*(Diagnostic Radiology)* See Rectifier

**Rectifier**  
*(Diagnostic Radiology)* Rectifiers convert alternating current (AC) to direct current (DC). The conversion process is known as rectification.

A rectifier is a diode (semiconductor or valve) or a combination of diodes with a specific arrangement that direct the current to pass only in one direction. The output of the rectifier is essentially half-AC current, which is then filtered into DC.

**Half-Wave Rectifiers:** Half-wave rectification uses either the positive or the negative AC wave. Half-wave rectifiers use either one diode (single-phase power supply) or three diodes (three-phase power supply). This type of rectification is relatively inefficient, because it blocks half of the input signal (Figure R.23).

**Full-Wave Rectifiers:** Full-wave rectifiers use both the negative and the positive waves of the AC signal, by inverting the polarity of one the waves. In case of single-phase power supply full-wave rectifiers use two diodes. One of them conducts during the positive AC wave, while the other one conducts during the negative AC wave (Figure R.24).

**Rectilinear scanner**  
*(Nuclear Medicine)* A scanner with a single collimated PM tube. The scintillation crystal used is sodium iodine (NaI (Tl)). The PM tube is connected, via electric circuitry to a light bulb. The intensity of the light bulb is proportional to the energy deposited. Beneath the light bulb is a very light sensitive film. The light bulb scans over the film in the same way as the PM tube scans over the patient, thus creating an image that represent tracer distribution in the patient.

The scanner was one of the first equipment used to image in vivo radionuclide distribution. But the scan is time consuming, one scan takes nearly 30 min and most of the rectilinear scanners have today been replaced by the scintillation camera invented by H. Anger in 1956.

Rectilinear scanners are also referred to as radioisotope scanners or scintigraphs.

**Related Article:** Radioisotope scanner

**Red, green, blue (RGB)**  
*(General)* RGB refers to the additive colour model where a wide range of colours is constructed from different combinations of the three primary colours red, green and blue. In any of colours in the RGB model is built up from the three colours and red, green and blue is referred to as the components of that specific colours. The RGB colour is determined by the individual intensity of components. When all the components have full intensity, the resulting colour is white, and when all components have zero intensity, the colour is black. A combination between two primary colours gives the secondary colours; magenta (red + blue), cyan (blue + green) and yellow (red + green). These together with black (key) constitute the CMYK (cyan, magenta, yellow, black) colour model, which is a subtractive colour model.

**Redistribution**  
*(Radiotherapy)* Redistribution, also called reassortment, is the return towards a more even distribution of cells within the cell cycle following the selective killing of those in the more radiosensitive phases at the time of irradiation.

A cell's susceptibility to radiation varies with the phase of the cell cycle: least sensitive to radiation late in the S phase and most sensitive in the M and G2 phases. Mammalian cell populations are asynchronous, that is cells are distributed throughout the cell cycle. The cells in the most sensitive phases will be killed preferentially by a dose of radiation. Therefore the majority of the surviving cells will be those in the more resistant phases resulting in a partial synchronisation of the cell population. If a fractionated radiotherapy regime is employed, the surviving cells will move through the cycle...
between fractions and some of the previously resistant cells will be in a more sensitive phase for the next delivery of radiation. This can be considered sensitisation resulting from redistribution and may result in a therapeutic gain since sensitisation by this mechanism only occurs in rapidly dividing cells and not in late-responding normal tissues. In practice, in human tumours the proportion of cells in the \(M\) and \(G_2\) phases is normally low, so the effect of redistribution is expected to be small. Additionally, although redistribution does not affect late-responding tissues it will occur in acutely responding normal tissues potentially limiting the therapeutic gain.

**Related Articles:** Cell cycle, Fractionation, Radiosensitivity, Repair, Repopulation, Reoxygenation, 5Rs of radiobiology


### Reducing agent

*(Nuclear Medicine)* The \(\text{Tc}^\text{m}\)-solution obtained from the \(\text{Mo}^{99}\text{Tc}^{\text{m}}\) generator is in the form of sodium pertechnetate with the oxidation state \(7^+\). This form is rather nonreactive and a lower oxidation state is needed to facilitate the labelling procedure. Various reducing agents have been used such as stannous chloride, stannous fluoride or stannous tartrate but stannous chloride is the most commonly used reducing agent. Technetium-99m is reduced to state \(4^+\) but also states \(3^+\) and \(5^+\) may be formed.


**Related Article:** Stannous chloride

### Reference air kerma rate (RAKR)

*(Radiotherapy, Brachytherapy)* Calibration of source strength is a very important part of a comprehensive brachytherapy quality system. The instruments, ion-chambers and electrometers, used for source strength determinations, should have calibrations that are traceable to national and international standards.

**Specification of Source Strength for Photon Emitting Sources:** Source strength for a photon emitting source can be given as a quantity describing the radioactivity contained in the source or as a quantity describing the output of the source:

1. Specification of contained activity
   a. Mass of radium; mg Ra
   b. Contained activity; Ci, Bq
2. Specification of output
   a. Equivalent mass of radium; mg Ra eq
   b. Apparent activity
   c. Reference exposure rate
   d. Reference air kerma rate
   e. Air kerma strength

The ICRU Report 38 – Dose and Volume Specification for Reporting Intracavitary Therapy in Gynecology – contains the following statement:

“It is recommended that radioactive sources be specified in terms of “reference air kerma rate”. The reference air kerma rate of a source is the kerma rate to air, in air, at a reference distance of 1 meter, corrected for air attenuation and scattering. For this purpose, the quantity is expressed in \(\mu\text{Gy*h}^{-1}\) at one metre’.

In modern brachytherapy dosimetry, reference air kerma rate or air kerma strength is the quantity used to calculate absorbed dose.

See *Source strength* for a full description of specification of source strength.

**Abbreviation:** ICRU = International Commission on Radiation Units and Measurements

**Related Articles:** Source strength, Mass of radium, Contained activity, Equivalent mass of radium, Apparent activity, Air kerma strength


### Reference depth

*(Radiotherapy)* The calibration of a radiation beam in external beam radiotherapy consists in establishing the relationship between the output of the equipment which produces the radiation and the output of the monitor which permits to determine the dose delivered to a patient. Calibration protocols have been developed for many years to measure the absorbed dose per monitor unit of an accelerator or the absorbed dose rate for a cobalt unit. The formalism applied for the determination of dose using the ionization chamber method is in principle the same in most dosimetry protocols. In addition to the choice of the ionisation chamber type, cylindrical or parallel plane, the protocols indicate the geometry of the measurements and the reference conditions. The proposed formalisms take into consideration the beam measurements performed at a reference depth which depends on the type of radiation and its energy. For photon beams the calibration itself should be performed through a measurement at the reference depth which is indicated in most dosimetry protocols as 5 or 10 \(g/cm^2\) depending on the beam quality. The dose at the depth of its maximum \(d_{\text{max}}\) can be calculated by dividing the dose measured at the reference depth by the appropriate percentage depth dose (PDD), tissue phantom ratios (TPR) or tissue maximum ratio (TMR) that are used for the clinical dosimetry. The choice of the protocols to calibrate the dose at a reference depth deeper than \(d_{\text{max}}\) is to avoid the influence by electrons scattered in the collimation system or in any other material in the beam. The dose at \(d_{\text{max}}\) is also dependent on the field size at high-energy photons. For electron beams the response of the ionization chamber per unit absorbed dose in a water phantom varies with the beam quality and the depth of measurement. This variation is determined mainly by the Spencer–Attix water to air stopping-power ratio \(\left( L/p \right)_{\text{air}} \). The expression is given by:

\[
\left( \frac{L}{p} \right)_{\text{air}} = \frac{\bar{L}}{\rho_{\text{air}}}
\]

at the chosen point of measurement. The beam quality and the reference depth in water must be specified in order to permit an accurate transfer of \(\left( L/p \right)_{\text{air}} \). In most protocols the first step is to determine the mean electron energy at the phantom surface \(E_0\)

\[
E_0 = \frac{1}{\bar{E}}
\]

using a given relationship between this energy and the half-value of the depth dose distribution \(R_{0.5}\). The chamber is then positioned at a specified reference depth \(d_{\text{ref}}\) in water and the value of \(\left( L/p \right)_{\text{air}}\) at \(d_{\text{ref}}\) is determined from tables which give \(\left( L/p \right)_{\text{air}}\) as a function of
$E_0$ and the depth in water. In most protocols the reference depth is normally chosen as $d_{\text{max}}$ or a specified depth which is deeper than $d_{\text{max}}$. For incident electron energies above 10 MeV the value of $d_{\text{max}}$ can vary by a large amount from machine to machine for beams of the same $R_{50}$. It follows that the value of $(E/p)_{\text{max}}$ at $d_{\text{max}}$ will also vary between machines for beams which have the same $R_{50}$. This leads to the need to express the Spencer–Attix water to air stopping-power ratio as a function of both $R_{50}$ and depth.

**Related Articles:** Percent depth dose (PDD), Tissue phantom ratio (TPR), Tissue maximum ratio (TMR), Stopping power ratio

**Reference ionisation chamber**  
(Radiation Protection) A reference ionisation chamber is a chamber that is used for the calibration of other chambers and it is designed for absolute dosimetry of photon, electron or proton beams. The specialised laboratory, for example the National Physics Laboratory in United Kingdom, calibrate and check dosimeters which are absolute standards. The secondary standards are the dosimeters used in hospitals and are regularly calibrated against the absolute standard. The substandards are the dosimeters calibrated against the secondary standard dosimeter.

In radiotherapy for absolute photon and electron dosimetry, the Farmer chamber (Figure R.25) is usually used. The Farmer chamber is cylindrical ionisation chamber with inner electrode made of aluminium and outer electrode of pure graphite. The insulation is made of teflon. The nominal volume is about 0.6 cm$^3$. The chamber is equipped with covers of different thicknesses to cover the outer electrode with the aim of achieving a constant calibration coefficient.

The Markus chamber (Figure R.26) is a parallel plate electron ion chamber applied for relative and absolute electron or proton dosimetry. The open air chamber, with a plate with a diameter of about 30 mm, depth of about 10 mm and nominal volume of 0.2 mm, is used to measure doses in water and solid type phantoms.

**Related Articles:** Cylindrical ionization chamber, Ionization chamber, Parallel plate ionization chamber, Thimble chamber  


**Reference isodose**  
(Radiotherapy, Brachytherapy) The reference isodose defines the reference volume, see the article Reference volume.

In the Paris dosimetry system, the reference isodose is 85% of the basal dose.  

**Related Article:** Reference volume

**Reference volume**  
(Radiotherapy, Brachytherapy)

*Reporting in Intracavitary Brachytherapy – ICRU Report 38:* The ICRU Report 38 (1985) defines the reference volume as 'the volume enclosed by the reference isodose surface. In order to facilitate intercomparisons between radiotherapy centres, it is necessary to agree upon a reference dose level. The treatment dose level defining the treatment volume may be equal to or different from this reference dose level. For reporting intracavitary therapy, it is necessary to determine the dimensions of the reference volume'. This means that the reference volume is used to compare different dose prescribing systems where different dose reference levels are used in centres.

The ICRU Report 38 recommends a reference dose level of 60 Gy. Reference dose levels on the order of 75–85 Gy are used for the high-risk target volume treated with a combination of external beam radiotherapy and brachytherapy.

*Reporting in Interstitial Brachytherapy – ICRU Report 58:* A reference volume is a volume encompassed by an isodose surface defined in relation to the mean central dose.

In the Paris system, the reference dose is 85% of the basal dose, and the basal dose is identical to the mean central dose in ICRU Report 58 (1997). The reference dose in the Paris system is also the minimum target dose of ICRU Report 58.

In the Manchester system for interstitial brachytherapy, the minimum target dose is about 90% of the prescribed dose.  

**Abbreviation:** ICRU = International Commission on Radiation Units and Measurements.

**Related Articles:** Paris system, Dosimetry systems, Reference isodose, ICRU reference point  

Reflection coefficient

(Ultrasound) When a propagating ultrasound wave encounters a medium with different acoustic impedance ($Z$) part of it will be reflected from the boundary, Figure R.27. The amplitude of the reflected beam will be higher if there is a large difference between the acoustic impedances. Acoustic impedance $Z$ can be defined as $Z = (\rho k)^{1/2}$ where $\rho$ is the tissue density and $k$ the stiffness.

Consider a material consisting of particles with mass $m$ linked together with springs with stiffness $k$, Figure R.28. If pressure is applied to one end, the particles are displaced. The movement will be easily transferred if the masses are small and the springs are weak. Conversely it is harder to move a material with large masses and stiff springs. As long as all the masses and springs are the same, the wave will be transferred with the same amplitude and particle velocity (assuming no damping effects). However, at the boundary between two materials with different $Z$ (different $\rho$ and/or $k$) these properties are changed for the transmitted wave and as either pressure or particle velocity can change abruptly this phenomenon implies that a second wave is formed at the boundary travelling in the opposite direction ($p_i = p_t + p_r$ and $v_i = v_t + v_r$). This is reflection.

Reflection back to the transducer plays an important role in the formation of ultrasound images as shown in Figure R.29. Reflection from smooth surfaces is described as specular reflection. This is a comparatively strong reflection and is very directionally dependent; echoes back to the transducer provide strong echoes contributing to the image; echoes in other directions do not. Reflections from irregular surfaces provide multi-directional echoes – those back in the direction of the transducer contribute to the image. In both cases, the onward-transmitted ultrasound energy is diminished.

The diagram on Figure R.29 depicts an ultrasound pulse meeting a boundary between two tissue types with different acoustic impedances. For specular reflection from a smooth surface the echo is reflected in one direction. If this is towards the transducer then the echo is detected and contributes to the image. If it is directed away from the transducer then the echo will not be detected. For a non-specular reflection, there are echoes in several directions – some is towards the transducer where it contributes to the image.

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Related Articles: Reflection coefficient, Acoustic impedance, Speed of sound
expressing the intensity reflection coefficient for plane waves are as follows: 

\[ R_i = I/I_0 = R_i^2, \quad R_i = ( (Z_2 - Z_1) (Z_1 + Z_2)^2 ) \] and 

\[ R_i = ( (Z_2 \cos \theta_i - Z_1 \cos \theta_i) / (Z_2 \cos \theta_i + Z_1 \cos \theta_i) ) \] which easily can be derived using the relationship \( I = p^2/2Z \) (see Intensity).

**Related Articles:** Acoustic impedance, Intensity, Snell's law, Transmission coefficient

**Refocusing**

*(Magnetic Resonance)* On application of an RF flip to a given volume magnetization of tissue a detectable RF transverse component appears. The transverse component is the vector sum of individual transverse component of spins throughout the volume. While all spin components remain in step or 'in phase', the detected signal will be a relatively strong oscillation at the Larmor frequency of the nuclei concerned. In practice, precession frequencies through any given volume will vary due to local variations in magnetic field strength, susceptibility effects and chemical shift. This range of precession frequencies causes spin transverse components to move apart from one another, dephasing the spins and causing signal to drop off.

Refocusing is the process of reversing this dephasing effect, causing signal to build up again towards a peak through the application of an RF refocusing pulse (see Figure R.30). The most effective refocusing is achieved through application of a 180° flip, although any flip angle will cause some degree of refocusing. Refocusing in a spin-echo type generates the spin echo at time TE, with the refocusing pulse applied at time TE/2 after the excitation pulse.

Refocusing can only 'correct' dephasing that is stable in time for any given spin. Magnet non-uniformities and susceptibility effects fall in this category, and refocusing will eliminate their dephasing effects. Dephasing caused by the intrinsic \( T_2 \) decay characteristics of the tissue is not stable in time and will not be rephased. For this reason the peak of an echo in a spin echo sequence lies on a \( T_2 \) decay curve.

**Refraction**

*(Ultrasound)* Refraction is the change in direction of a wave due to its speed. In medical ultrasound this occurs when the sound wave crosses the boundary of two tissues with different speeds of sound.

**Related Articles:** Snell's law, Speed of sound, Reflection coefficient

**Region of interest (ROI)**

*(Nuclear Medicine)* The region of interest is a user defined region which is commonly abbreviated as ROI. In nuclear medicine...
Registration

ROIs are often placed around particular regions or organs when evaluating the organ functionality, radiocompound kinetics, activity quantification, etc. When performing quality controls or when validating the camera parameters of a new emission image system the ROI can be used to measure the spatial resolution, uniformity, etc.

An example of a practical implementation of the ROI is when measurement is made of the spatial resolution in a scintillation camera. The first step is to image a line source. An ROI placed over the line source will produce a source profile (which for a number of reasons is smeared out). A measure of the spatial resolution is the full width at half-maximum of the source profile.


**Regulatory authority**

*(Radiation Protection)* The regulatory authority is the national authority designated by the government that regulates the introduction and conduct of any practice involving sources of radiation. The regulatory authority should be independent of any governments departments and agencies that are responsible for the promotion and development of the practices being regulated. The regulatory authority must also be independent of registrants, licensees and the designers and constructors of the radiation sources used in practices.

In some counties regulatory responsibility for different practices or different aspects of radiation safety may be divided between different authorities.


**Reject film analysis**

*(Diagnostic Radiology)* Reject film analysis is a procedure used in radiography for quality assurance. The procedure is to collect the films that have been rejected (and usually repeated), review and evaluate them to determine reason for rejection, and then sort and tabulate by causes such as patient positioning, technique error, processing, etc. The rejects can also be tabulated with respect to individual radiographers who performed the examination.

The results of the analysis are used to identify the significant causes for rejecting films so that specific actions (training, equipment calibration, revised procedures, etc.) can be taken to reduce rejects in the future.

**Rejection method**

*(General)* Occasionally the distribution function method is cumbersome to use due to mathematical difficulties in the calculation of the inverse of the cumulative probability distribution function (CPDF). One can use the rejection method described by the following three steps to obtain a random sample:

1. Let the probability distribution function, \( \text{pdf}(x) \), be bounded in the range \([a, b]\). Calculate a normalized function \( \text{pdf}^*(x) = \text{pdf}(x) / \max \{\text{pdf}(x)\} \) so the maximum value of \( \text{pdf}^* \) is equal to unity.
2. Sample a uniform distributed value of \( x \) within the range \([a, b]\) from the relation \( x = a + R \cdot (b/a) \) and where \( R \) is a random number.
3. A second random number \( R_2 \) with then decide whether the sampled \( x \) should be accepted. This choice is made by calculating the function value of \( \text{pdf}^*(x) \) from the sampled \( x \) value and then check if \( R_2 < \text{pdf}^*(x) \). If this relation is fulfilled, then \( x \) is accepted as a proper distributed stochastic value. Otherwise, a new \( x \) value needs to be sampled, according to the procedure in step 2.

Related Articles: Monte Carlo, Distribution function method

**Relative anisotropy (RA)**

*(Magnetic Resonance)* The relative anisotropy (RA) is a diffusion anisotropy index, determined from the elements of the diffusion

![Effect of refraction on beam direction](Image)

**FIGURE R.33** Refraction through bone can lead to large changes in beam direction with corresponding loss of energy deep to the site of refraction.

Related Article: Image fusion

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Related Articles: Monte Carlo, Distribution function method
tensor. Commonly, the scaled relative anisotropy (sRA) is calculated, with the sRA defined as

\[
sRA = \sqrt{\frac{(\lambda_i - \bar{\lambda})^2 + (\lambda_j - \bar{\lambda})^2 + (\lambda_k - \bar{\lambda})^2}{6\bar{\lambda}^2}} = \sqrt{\frac{\sum (\lambda^2 - \bar{\lambda})}{6\bar{\lambda}^2}}
\]

where

- \(\lambda_i\) is the \(i\)th eigenvalue of the diffusion tensor
- \(\bar{\lambda}\) is the mean of the eigenvalues, that is the mean ADC

The sRA ranges from 0 (isotropic diffusion) to 1 (anisotropic diffusion) and is related to RA as sRA \(\times 2^{1/2} = RA\). High RA values are found in white matter, which in which the diffusion is highly dependent on the direction in which the diffusion is studies. Lower RA values are found in grey matter, where the rate of diffusion is more similar in all directions.

The relative anisotropy is a rotationally invariant metric, that is it is independent of the major diffusion direction.

**Related Article:** Diffusion tensor

**Relative biological effectiveness (RBE)**

(Radiation Protection; General) Relative biological effectiveness (RBE) accounts for the varying destructive effects of different radiations on tissue. RBE is expressed relative to x- or gamma-ray photons, so the RBE of these radiations is unity. Beta radiation also has an RBE = 1.

Studies have shown that, for equal absorbed doses, alpha, proton, neutron and other ionising radiation is more destructive than photon or beta radiation. Proton radiation is known to be around 2 times more destructive than photons, alpha radiation is known to be between 4 and 16 times more destructive than photons, and neutrons are up to 8 times more destructive.

Effect on tissue is also dependent on other aspects of the radiation, for example energy and radiation quality (spectral characteristics). For these reasons, and due to incomplete knowledge of the risk levels for all kinds of radiations at all energies and different qualities, it was decided to adopt single values of 2 for protons, 20 for alpha particles and other heavy fission products, and use an equation modelling a continuous curve for neutrons:

\[
w_{RBE} = \begin{cases} 
2.5 + 18.2 \exp\left[\frac{-(\ln E_s)^2}{6}\right] & \text{for } E_s < 1 \text{ MeV} \\
5.0 + 17.0 \exp\left[\frac{-(\ln(2E_s))^2}{6}\right] & \text{for } E_s \geq 1 \text{ MeV}
\end{cases}
\]

The radiation-weighted or equivalent dose is then calculated as \(H = w_{RBE} \times D\) for an absorbed dose \(D\). The adoption of over-cautious values of \(w_{RBE}\) allows general risk estimates to incorporate radiation factors. More specific radiation-weighting factors are used in radiotherapy to accurately calculate doses.

**Related Articles:** Equivalent dose, Radiation weighting factor

**Further Reading:** Relative Biological Effectiveness (RBE), Quality Factor (Q), and Radiation Weighting Factor (\(w_{RBE}\)), ICRP Publication 92 Ann. ICRP 33(4), 2003.

**Relative biological effectiveness (RBE)**

(Radiotherapy) Relative biological effectiveness (RBE) is the ratio of dose at a reference radiation quality and dose of a test radiation that produces an equal biological effect. The reference radiation is usually 250kV x-rays, but should always be specified.

The response of tumours and normal tissues to radiation is affected by the type of radiation. As the linear energy transfer (LET) increases, the radiation produces more cell killing per Gy. Figure R.34 shows a typical survival curve for cells exposed to x-rays and fast neutrons.

The initial shoulder for fast neutrons is smaller than that for x-rays and the final slope is steeper. This indicates that for high-LET radiation either there is a higher ratio of lethal to potentially lethal lesions or that radiation damage is less likely to be repaired correctly.

The RBE depends on the LET of the radiation, the radiation dose, the number of dose fractions, the dose rate and the biological system or end-point.

Figure R.35 illustrates the RBE dependence on LET. The positions of the maxima have been measured for a range of mammalian cells and found to be at LET values in the region of 100keV/\(\mu m\). This maximum value of RBE can be understood in terms of the production of double-strand DNA breaks. For low-LET radiations, for example x-rays, there is a low probability that a single track will cause a double-strand break and therefore the biological effectiveness of the radiation is low. For very high-LET radiations, for example those with LET of 200keV/\(\mu m\), there is a high probability that a single track will result in a double-strand break but since the ionising events occur very close together, more energy is deposited than is necessary. This effect is sometimes referred to as overkill.

**Figure R.34** Typical survival curve for cells exposed to x-rays (blue) and fast neutrons (red).

**Figure R.35** RBE increases with LET, reaching a maximum at a LET of around 100keV/\(\mu m\) beyond which it falls. Radiation with a LET of 100keV/\(\mu m\) is therefore optimal in terms of producing a biological effect. Note that the LET scale is a log scale.
Since RBE is the ratio of dose producing equal biological effect this very high-LET radiation has a lower RBE than the optimal LET radiation. The optimal LET radiation is that for which the average separation of ionising events is on the order of the DNA diameter. Examples of optimal LET radiations include low-energy protons and alpha-particles.

RBE depends on the radiation dose and the number of fractions because the shape of the dose–response relationship varies for radiations with substantially different values of LET as illustrated in Figure R.36.

Dose rate can affect RBE because the dose–response curve for low-LET radiations varies with dose rate (see the article on Dose rate dependence) but there is little effect on that for high-LET radiation.

In general, RBE values are high for biological systems that accumulate and repair a substantial amount of sub-lethal damage and low for those that do not. A more detailed discussion of RBE can be found in the book by Hall and Giaccia.

**Abbreviations:** LET = Linear energy transfer and RBE = Relative biological effectiveness.

**Related Articles:** Cell survival curve, Dose rate dependence, Dose–response model, Fractions, Fractionation, Linear energy transfer, Neutron therapy, Radiation quality, Repair


**Relative electron density**

(Radiotherapy) Relative electron density (RED) for some material denotes the electron density (i.e. the number of electrons per volume unit) for that material divided by the electron density for water. In the following table, relative electron densities for some tissues are reported.

<table>
<thead>
<tr>
<th>Material</th>
<th>Physical Density (g/cm³)</th>
<th>Electron Density per cm³ x 10^21</th>
<th>Electron Density Relative to Water (RED)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Water</td>
<td>1.00</td>
<td>3.340</td>
<td>1.000</td>
</tr>
<tr>
<td>Lung (inhale)</td>
<td>0.20</td>
<td>0.634</td>
<td>0.190</td>
</tr>
<tr>
<td>Lung (exhale)</td>
<td>0.50</td>
<td>1.632</td>
<td>0.489</td>
</tr>
<tr>
<td>Adipose</td>
<td>0.96</td>
<td>3.170</td>
<td>0.949</td>
</tr>
<tr>
<td>Breast (50/50)</td>
<td>0.99</td>
<td>3.261</td>
<td>0.976</td>
</tr>
<tr>
<td>Muscle</td>
<td>1.06</td>
<td>3.483</td>
<td>1.043</td>
</tr>
<tr>
<td>Liver</td>
<td>1.07</td>
<td>3.516</td>
<td>1.052</td>
</tr>
<tr>
<td>Trabecular bone</td>
<td>1.16</td>
<td>3.730</td>
<td>1.117</td>
</tr>
<tr>
<td>Dense bone (800mg/cc)</td>
<td>1.61</td>
<td>5.052</td>
<td>1.512</td>
</tr>
<tr>
<td>Dense bone (1000mg/cc)</td>
<td>1.66</td>
<td>5.243</td>
<td>1.570</td>
</tr>
<tr>
<td>Dense bone (1250mg/cc)</td>
<td>1.83</td>
<td>5.718</td>
<td>1.712</td>
</tr>
<tr>
<td>Dense bone (1500mg/cc)</td>
<td>2.00</td>
<td>6.209</td>
<td>1.859</td>
</tr>
<tr>
<td>Dense bone (1750mg/cc)</td>
<td>2.17</td>
<td>6.698</td>
<td>2.005</td>
</tr>
<tr>
<td>Titanium</td>
<td>4.51</td>
<td>12.475</td>
<td>3.735</td>
</tr>
</tbody>
</table>

**Related Article:** Electron density

**Relative humidity**

(General) Relative humidity describes the amount of water vapour that exists in a gaseous mixture of air and water. Relative humidity (RH) is defined as the ratio of actual vapour density to saturation vapour density at a given temperature. RH therefore indicates the maximum amount of vapour that air can hold at a given temperature. The quantitative expression for relative humidity as percentage is given by

\[
\% \text{Relative humidity} = \left( \frac{\text{Vapour density}}{\text{Saturation vapour density}} \right) \times 100
\]

The partial pressure of water vapour is proportional to its concentration and so %RH can be expressed in terms of vapour pressure which represents the partial vapour pressure contributed by the water molecules. Therefore %RH is also given by the ratio of the actual partial pressure to the saturated partial pressure of water vapour at a prescribed temperature (Figure R.36).

**Relative risk**

(Radiation Protection) Relative risk is used to compare the risk from different factors in causing harm (such as medical x-rays and smoking cigarettes), including late effects such as cancer, in an exposed population. It can also mean the comparison (the ratio) of the risk of harm between two populations where one is exposed to a factor, and the other group is not.

Relative risk should always be analysed together with the total or absolute risk.

**Related Articles:** Absolute risk, Excess risk

**Relaxation**

(Magnetic Resonance) In magnetic resonance imaging (MRI) the term ‘relaxation’ refers to different processes by which nuclear magnetization in a non-equilibrium state (typically created by radiofrequency excitation) returns to the equilibrium distribution. Different physical processes cause different rates of spin relaxation in different directions with respect to the static magnetic field \( B_0 \). Relaxation is typically divided into two types, that is spin–lattice relaxation (longitudinal relaxation, \( T_1 \) relaxation) and spin–spin relaxation (transverse relaxation, \( T_2 \) relaxation). The rates of spin–lattice and spin–spin relaxation are described by the time constants \( T_1 \) and \( T_2 \), respectively, often referred to as relaxation times. Note that in an ideal environment, where the nucleus being observed is completely isolated, relaxation would not exist. In such an idealised environment, magnetisation which is set into a non-equilibrium state cannot equilibrate.
Spin–Lattice Relaxation (Longitudinal Relaxation, $T_1$ Relaxation): Spin–lattice relaxation is the process during which the spins dispose of the energy that was obtained from the radiofrequency (RF) pulse during excitation. The energy is transferred back to the surrounding physical and chemical environment (i.e. the lattice) in order to restore the equilibrium state. The lattice consists of neighbouring nuclei or molecules that show vibrational, rotational or translational motion, forming a local magnetic field called the lattice field ($B_L$). The correlation time $\tau_c$ (i.e. the average time between collisions) is often used to describe the mobility of the molecules. An environment with long $\tau_c$ is characterized by slower movements and lower frequencies, while short $\tau_c$ corresponds to higher mobility and higher frequencies. In biological tissue, water is assumed to exist in a number of binding states, from tightly bound water molecules (long $\tau_c$ and low frequency) at the surface of macromolecules (such as proteins and polysaccharides) via a ‘structured’ component (medium $\tau_c$), at an intermediate distance from the large molecule, to more or less free water (short $\tau_c$ and high-frequency components) farthest away from the macromolecule. The complex motion patterns lead to fluctuations in the lattice field that correspond to the resonance frequency $\omega_0$ of the spins will enable interaction with excited nuclei, causing them to return to the lower-energy state. The energy that is released by the nucleus will increase the lattice vibrations and rotations (i.e. transfer into heat). Note that the excess energy is not emitted as radiation. Since the spectral density function of tissue is not uniform over all frequencies, the spin–lattice relaxation (and consequently $T_1$) tends to be dependent on the magnetic field strength. In tissue, an increased magnetic field strength (implying a higher resonance frequency) leads to prolonged $T_1$ due to fewer components of $B_R$ fluctuations at higher frequencies. Formally, the longitudinal relaxation time $T_1$ is the time constant for the recovery of the component of the magnetization vector $M$ that is parallel to the main magnetic field $B_0$, denoted $M_z$. The increase of $M_z$ due to longitudinal relaxation is thus described by

$$M_z(t) = M_z(0) \cdot 1 - e^{-t/T_1}$$

where

- $M_z(0)$ is the equilibrium magnetization (along the z-axis)
- $t$ is time

Different tissues show different $T_1$ values (see Relaxation time).

Spin–Spin Relaxation (Transverse Relaxation, $T_2$ Relaxation): Spin–spin relaxation refers to interaction between excited nuclei, leading to dispersion of transverse magnetization that is out of equilibrium. The spin system does not lose any energy in the spin–spin relaxation process, but the phase coherence of spin precession is gradually lost. The phase dispersion is caused by inhomogeneities in the local static magnetic field, and only the inhomogeneities internal to the proton system contribute to true $T_2$ relaxation. Free water molecules with short $\tau_c$ display rapid movements, and one particular magnetic dipole will thus perceive high-frequency fluctuations in the local magnetic field, effectively averaging out over a few milliseconds. Due to this so-called motional averaging the spins experience a relatively homogeneous local field and limited dephasing. Bound water molecules (close to macromolecules), on the other hand, display large low-frequency motion components (close to zero frequency) and this corresponds to static-field fluctuations, that is a local magnetic field inhomogeneity. Obviously, such an environment will lead to substantial phase dispersion and an efficient spin–spin relaxation. With regard to the spectral density function, $T_2$ relaxation is governed by the motion components close to zero frequency (corresponding to fluctuations in the static magnetic field). The transverse relaxation time $T_2$ is the time constant for the decay of the component of the magnetization vector $M$ that is perpendicular to the main magnetic field $B_0$ after RF excitation, denoted $M_{xy}$. The decrease of $M_{xy}$ due to transverse relaxation is thus described by

$$M_{xy}(t) = M_{xy}(0) \cdot e^{-t/T_2}$$

where

- $M_{xy}(0)$ is the transverse magnetization immediately after RF excitation
- $t$ is time

Different tissues show different $T_2$ values (see Relaxation time), but $T_2$ does not display any pronounced dependence on the magnetic field strength within the range of $B_0$ values typically used in MRI.

$T_2^*$ and the Static Magnetic Field Inhomogeneity: Inhomogeneities in the static magnetic field of an MRI unit are an external source of phase dispersion similar to the effects of spin–spin relaxation. The phase dispersion caused by static field inhomogeneities is, however, not a true relaxation process. The inhomogeneities are static in time and dependent on the location of the spin in the magnet, that is the signal component lost due to magnetic field inhomogeneities can be recovered by performing a spin echo experiment. $T_2^*$ is the time constant for the decay of $M_{xy}$ when effects of true $T_2$ relaxation as well as static field inhomogeneities are included:

$$\frac{1}{T_2^*} = \frac{1}{T_2} + \gamma \Delta B_0$$

where

- $\gamma$ represents the gyromagnetic ratio
- $\Delta B_0$ is the static magnetic field inhomogeneity (i.e. the local variation in $B_0$)

The $T_2^*$ relaxation time is always shorter than the $T_1$ relaxation time.

Related Articles: Relaxation time, Relaxation rate

**Relaxation rate**  
(Magnetic Resonance) In MRI, the relaxation rate (in s⁻¹) is defined as the inverse of the relaxation time, for example the longitudinal relaxation rate $R_1$ is given by $R_1 = 1/T_1$ and the transverse relaxation rate $R_2$ is given by $R_2 = 1/T_2$.

**Related Articles:** Relaxation, Relaxation time

**Relaxation time**  
(Magnetic Resonance) In MRI, the term ‘relaxation time’ is used for the time constants associated with changes in the magnetization vector $M$ after excitation of the spin system with one of more radio-frequency (RF) pulses.

**Spin–Lattice Relaxation Time $T_1$:** The spin–lattice or longitudinal relaxation time $T_1$ is the time constant for the recovery of the component of the magnetization vector $M$ that is parallel to the main magnetic field $B_0$, denoted $M_z$. The increase of $M_z$ due to longitudinal relaxation is described by

$$M_z(t) = M_0 \left(1 - e^{-t/T_1}\right)$$

where $M_0$ is the equilibrium magnetization (along the $z$-axis) $t$ is time

$T_1$ displays a dependence on the magnetic field strength, and different tissues show different $T_1$ values (see Table R.1).

**Spin–Spin Relaxation Time $T_2$:** The spin–spin or transverse relaxation time $T_2$ is the time constant for the decay of the component of the magnetization vector $M$ that is perpendicular to the main magnetic field $B_0$, denoted $M_{xy}$. The decrease of $M_{xy}$ due to transverse relaxation is described by

$$M_{xy}(t) = M_{xy}(0) \cdot e^{-t/T_2}$$

where $M_{xy}(0)$ is the transverse magnetization immediately after RF excitation $t$ is time

Different tissues show different $T_2$ values (see Table R.1).

<table>
<thead>
<tr>
<th>Tissue</th>
<th>$T_1$ (ms)</th>
<th>$T_2$ (ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Grey matter</td>
<td>800</td>
<td>1000</td>
</tr>
<tr>
<td>White matter</td>
<td>500</td>
<td>600</td>
</tr>
<tr>
<td>Muscle</td>
<td>550</td>
<td>900</td>
</tr>
<tr>
<td>Fat</td>
<td>200</td>
<td>250</td>
</tr>
</tbody>
</table>

*Note:* Relaxation-time data vary considerable between different literature sources. More detailed information is provided in standard MRI textbooks (e.g. McRobbie et al. and references therein).

**$T_2^*$ Relaxation Time and the Static Magnetic Field Inhomogeneity:** $T_2^*$ is the time constant for the decay of $M_{xy}$ when effects of true $T_2$ relaxation as well as static field inhomogeneities are included:

$$\frac{1}{T_2} = \frac{1}{T_2^*} + \gamma \Delta B_0$$

where $\gamma$ represents the gyromagnetic ratio $\Delta B_0$ is the static magnetic field inhomogeneity (i.e. the local variation in $B_0$)

The $T_2^*$ relaxation time is always shorter than the $T_2$ relaxation time.

**Related Articles:** Relaxation, Relaxation rate


**Relaxivity**  
(Magnetic Resonance) The relaxivity $r$ describes the ability of a chemical compound (often a contrast agent) to increase the relaxation rate $R$ of the surrounding proton spins. Relaxation is divided into longitudinal relaxation ($T_1$ relaxation) and transverse relaxation ($T_2$ relaxation), and the corresponding $T_1$ and $T_2$ relaxivities are denoted $r_1$ and $r_2$, respectively. The relaxivity $r_i$ ($i = 1, 2$) is defined as the change in relaxation rate $R_i$ per unit concentration of contrast agent and is typically expressed in units of $m/M/s$. The relaxivity is normally defined as the slope of the linear-regression equation obtained from a plot of measured relaxation rate versus the concentration $c$ of the contrast agent:

$$R_i = R_{i0} + r_i c$$

where $R_{i0}$ is the relaxation rate of the solvent without contrast agent. Relaxivities generally depend on a number of factors, such as magnetic field strength, temperature and physiological environment.

**Related Articles:** Relaxation, Relaxation rate, Relaxation time


**Relaxometry**  
(Magnetic Resonance) The most common magnetic resonance imaging (MRI) techniques for a quantitative diagnosis are relaxometry (R), magnetization transfer (MT), diffusion imaging (DWI), volumetry and spectroscopy (MRS). Relaxometry refers to the study and/or the measurement of the relaxation variables in MRI and their dependence on physical parameters. One can create a map, based on the relaxation time itself. Generally, applications for the characterization of tissues involve $T_2$ and $T_2^*$, making use of spin-echo sequences with two or more different echo times (TE) and a long repetition time (TR). $T_2$ and $T_2^*$ relaxometry maps may be generated either by spin-echo or by gradient echo sequences, with the latter being used to measure $T_2^*$. Consequently the results may be noisier because the influence of inhomogeneities in the magnetizing field is greater. At least two images are needed to generate a map of...
A typical installation of a remote afterloading unit for high dose rate brachytherapy:

1. In the treatment room: The treatment unit, the trolley, with a shielded safe
   a. One single source
      i. Geometrically small
      ii. High specific source strength (e.g. Iridium 192, ‘10 Ci’)
   b. Stepping source movement, requiring precision drive motor
   c. Computer controlled source movement of
      i. Dwell time
      ii. Step size
   d. Several treatment channels – computer controlled indexing between channels
   e. Applicators and source guide tubes, transfer tubes, to connect applicators to the trolley head (applicators must be closed – the source must not come into contact with body fluids)
   f. Built in safety systems
      i. Dummy source (not radioactive) for verification of applicator connections (identical to the active source)
      ii. Detector for radiation level measurements (indication on control panel)
      iii. Hand crank to rewind the source cable manually
   g. Connection to the treatment console and the treatment planning system
2. In the control room: The treatment console and computer
   a. Console
      i. Start and control of the treatment (interrupt, emergency stop)
      ii. Indicators for the treatment
   b. Computer with printer
      i. Treatment control software/source movements
      ii. Documentation of treatment
      iii. Connection to trolley
      iv. Connection to treatment planning system
3. More pieces of equipment
   a. Patient monitoring systems
      i. Two-way audio system (intercom)
      ii. Television system
   b. Door interlock system
   c. Emergency container in the treatment room
   d. Independent radiation level monitor in the treatment room with separate battery back-up; radiation level ‘high’ indicated both by sound and light
   e. Radiation survey meter

Remote afterloading:

1. Applicators, needles, catheters, etc. are inserted.
2. Correct positions are verified using dummy sources.
3. The source is loaded into the applicator/s using a remote controlled afterloading unit. Relevant personnel operate the afterloading unit from the operator’s room close to the treatment room (compare to ‘linac’ treatments).

Remote afterloading unit

(Radiotherapy, Brachytherapy) See Remote afterloading unit

Source Handling and Loading: The brachytherapy source/s must be handled and loaded into the applicators for treatment, and many methods have been used over the time. These methods have been developed primarily to reduce the dose to the personnel but also to improve the quality of the treatment itself.

Related Articles: Photoelectric relay, Delay relay

FIGURE R.38  Electromagnetic relays in older type of x-ray equipment.
Reorientation

(Nuclear Medicine) Initially reconstructed myocardial perfusion images are aligned with patient coordinates as transversal images.

Reoxygenation

(Radiotherapy) Reoxygenation is the process by which hypoxic clonogenic cells become better oxygenated during the period after irradiation.

Over the years, many chemical and pharmacological agents have been discovered that modify the response of mammalian cells to radiation. The simplest of these, and the one that possibly produces the most dramatic effect, is oxygen. It is known that tumour cells can be hypoxic and therefore less sensitive to radiation that indirectly damage DNA through free radicals produced by the ionisation of oxygen. Therefore, the presence or absence of oxygen dramatically influences the biological effect of sparsely ionising radiations such as x-rays but there is no effect for densely ionising radiations such as α-particles.

The oxygenation status of human tumours has been determined with a variety of techniques including measuring the distance between tumour cells and vessels in histological sections, determining the oxygen saturation of haemoglobin, monitoring changes in tumour metabolism through to the newer techniques of oxygen probes, hypoxic markers, the comet assay and non-invasive imaging. Hypoxia has been demonstrated as a common feature of human solid tumours that can influence both the malignant progression and the response of tumours to radiotherapy. However, it can result from two quite different mechanisms. Chronic hypoxia results from the limited diffusion distance of oxygen through tissue that is respiring. The distance to which oxygen can diffuse is largely limited by the rapid rate at which it is metabolised by respiring tumour cells. In contrast, acute hypoxia is the result of the temporary closing of a tumour blood vessel owing to the malformed vasculature of the tumour. There is good evidence that tumour blood vessels open and close in a random fashion and so different regions of the tumour become hypoxic intermittently. At the moment of irradiation, a proportion of cells may be hypoxic but if the radiation is delayed to another point in time, a different group of cells may be hypoxic. The oxygen status of cells in a tumour is dynamic and constantly changing.

In the 1960s van Putten and Kallman performed experiments using a transplantable sarcoma in a mouse and demonstrated that during a course of treatment hypoxic cells become oxygenated. They found that the fraction of hypoxic cells in a tumour is about the same at the end of a fractionated regime of radiotherapy as in the untreated tumour. If the hypoxic cells were not reoxygenated...
during the course of treatment, the proportion of hypoxic cells would increase since the radiation depopulates the aerated-cell population more than the hypoxic-cell population.

In animal experiments, it has been found that some tumours take several days to reoxygenate, in others the process is complete within just a few hours, and in some tumours both fast and slow components of reoxygenation are evident. The type of hypoxia being reversed, chronic or acute, is reflected in the differences in time scale. The slow component stems from the reoxygenation of chronically hypoxic cells that occurs as the tumour shrinks: cells that were beyond the range of oxygen diffusion are closer to a blood supply. The fast component stems from the acutely hypoxic cells reoxygenating as tumour blood vessels open and close: cells that were hypoxic at the time of irradiation because they were in regions where a blood vessel was temporarily closed reoxygenate quickly when that vessel is opened.

Clearly, reoxygenation of tumours has important implications for radiotherapy. If human tumours do reoxygenate as rapidly and efficiently as most of the animal tumours that have been studied, then the use of a multi-fraction course of radiotherapy may be all that is required to deal effectively with any hypoxic cells. Unfortunately, knowledge of the time course of reoxygenation in human tumours is not known and in fact it is not known with certainty whether any do actually reoxygenate. However, supporting evidence exists from the standard clinical use of fractionated treatments where many tumours are eradicated with doses on the order of 60 Gy given in 30 fractions which would be unlikely in the presence of a very small proportion of hypoxic cells.

**Related Articles:** Alpha beta ratio, Fractionation, Interruption of treatment, Radiosensitivity, Repair, Redistribution, Repopulation, 5Rs of radiobiology


**Repair (Radiotherapy)** Repair refers to the process by which the function of macromolecules is restored following radiation damage. As a result there is an increase in cell survival or a reduction in the extent of radiation damage to a tissue when time is allowed for repair to occur.

DNA is considered the principal target for the biological effects of radiation, including cell killing, carcinogenesis and mutation. DNA is a large molecule with a double helix structure consisting of two strands. If cells are irradiated with a modest dose of x-rays, many breaks of a single strand (SSB) may occur. Such lesions are of little biological consequence for cell killing since they are repaired easily using the opposite strand as a template. Likewise, if both strands are broken but the breaks are far apart, repair again can occur readily since the two breaks are handled separately. However, if the break in the two strands are opposite one another (or within a few base pairs), a double strand break (DSB) may result. DSBs are regarded as the most important radiation-induced lesions.

Radiation damage to mammalian cells can be categorised as

1. **Lethal damage (LD):** Irreversible and irreparable, leading irrevocably to cell death.
2. **Potential lethal damage (PLD):** Radiation damage that under normal circumstances causes cell death but can be modified by post-irradiation environmental conditions.
3. **Sublethal damage (SLD):** Damage which under normal circumstances can be repaired within hours unless additional sublethal damage is added, for example a second dose of radiation, with which it can interact to form lethal damage.

4. **Nonlethal damage (NLD):** It results in cells with heritable lesions (sometimes called lethal mutations) which do not prevent proliferation but may affect the rate of proliferation.

Repair is possible for PLD and SLD.

PLD is repaired if post-irradiation conditions are sub-optimal for growth. In these circumstances, mitosis is delayed so there is the potential for any damage to the chromosomes to be repaired before it is attempted. Repair of PLD has been observed in transplantable animal tumours so it is reasonable to suppose that it also occurs in human tumours. Indeed, it has been suggested that radioresistant human tumours may have more efficient mechanisms to repair PLD than radiosensitive tumours. This hypothesis has yet to be proven.

The effect of a given dose of radiation is less if it is split into two fractions delivered a few hours apart. This is termed repair of SLD. Considerable repair occurs within 15–60 min with complete recovery usually by around 4–6 h, although repair seems to be slower in some normal tissues such as the spinal cord. The effect of the repair of SLD can be seen in the cell survival curve of a cell population receiving a dose split into two equal doses separated by a time interval compared with that of a cell population receiving the whole dose in a single fraction, Figure R.42. More cells survive in the population that receive the fractionated treatment than for the single fraction population because the shoulder of the curve must be repeated with each fraction. In general, there is a good correlation between the extent of sublethal damage repair and the size of the shoulder of the cell survival curve.

In terms of the linear quadratic model description of the survival curve, tissues with small values of the $\alpha/\beta$ ratio have high sublethal damage repair potential while repair potential is limited for tissues with large values of $\alpha/\beta$. Hence, tissues with low $\alpha/\beta$-ratios are sensitive to the dose fractionation scheme (fractionation effect) while those with high $\alpha/\beta$-ratios are hardly affected by dose fractionation. This has implications for radiotherapy treatment since tumours generally have high values of the $\alpha/\beta$-ratio while late responding normal tissues have low values of the $\alpha/\beta$-ratio. Therefore a fractionated regime may enhance the therapeutic effect.

**Abbreviations:** DNA = Deoxyribonucleic acid, DSB = Double strand break, LD = Lethal damage, NLD = Non-lethal damage, PLD = Potential lethal damage, SLD = Sublethal damage and SSB = Single strand break.

**Related Articles:** Alpha beta ratio, Fractionation, Linear quadratic (LQ) model, Linear-quadratic dose–response curve, Radiosensitivity, Repopulation, Redistribution, Reoxygenation, 5Rs of radiobiology, Surviving fraction, Therapeutic effect.


**Repair of radiation damage (Radiation Protection)** When cells are exposed to ionising radiation, damage can occur, both in the DNA itself, and in other biological molecules within the nucleus and cytoplasm. This damage does not necessarily lead to adverse effects, or bioeffects. Cells have proteins and enzymes whose main function is to act as part of the mechanisms to repair damage to DNA. Ordinarily these mechanisms work very well to repair radiation damage.

For more information, see the articles on Bioeffects, Radiation damage and Radiobiological models.

**Related Articles:** Bioeffects, Adverse effects, Radiation damage, Radiobiological models.
**Repetition time (TR)**

(Repetition time (TR) is the time between successive pulse sequence executions, applied to the same slice or volume of interest. In standard MR imaging, a pulse sequence is repeated in order to achieve sufficient k-space coverage and the phase encoding gradient is changed one (or several) times during each execution. In single-shot sequences (acquiring the whole k-space information during one repetition), the term ‘TR’ is still used if several images of the same volume are acquired in a time series, for example in fMRI applications.

TR, echo time (TE) and excitation pulse flip angle are important parameters in the build-up of image contrast. As an example, a very long TR (approximately five $T_1$ relaxation times) allows full build-up of the longitudinal magnetization to the order of its thermal equilibrium value after excitation, while a shorter TR does not. In the latter case, different $T_1$ values in different tissue types give different build-up of longitudinal magnetization during TR, and when the magnetization is repeatedly flipped into the transverse plane, the detected signal will contain $T_1$ contrast information (Figure R.43).

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**Related Articles:** Echo time (TE), Encoding gradients, Flip angle, Relaxation time

**Further Readings:**


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**Repopulation**

(Repopulation refers to the process following irradiation whereby the number of cells in a normal tissue or tumour is restored by the proliferation of surviving cells.

For fractionated radiotherapy treatment, ideally the normal tissue would be completely repopulated following irradiation while the tumour would show no growth between fractions. If this were the case, the tumour would be progressively depopulated while the surrounding normal tissue would maintain a steady state. There is in general no difference in cell cycle times between tumour cells and those of normal tissues but their effective doubling times may show considerable differences due to the effects of cell loss and other factors influencing cell population kinetics.

In some normal tissues there may be a large increase in the proliferation rate following the initial radiation injury until the normal tissue is fully reconstituted. This is seen, for example in haemopoietic stem cells which show an immediate response with both an increase in the growth fraction and a decrease in the maturation rate. Studies of skin damage in mice show that the effects of repopulation are seen at around 12 days into a fractionated regime after which the extra dose required to compensate for proliferation increases very rapidly with time. The behaviour of such early-responding normal tissues is considerably different from that of late-responding tissues such as spinal cord. Studies of spinal cord in rats show that the effects of repopulation occur much later in a fractionated regime (40 days compared with the 12 days observed for skin in mice) with a slower increase in the extra dose required to compensate for proliferation with time. Comparable data for humans are not available and the timescales are expected to be very much longer than those observed in rodents. However, if the experimental data on rodents were to be translated to clinical radiotherapy for humans, it would be reasonable to assume that for conventional radiotherapy protocols, early-responding tissues are triggered to proliferate within a few weeks of the start of fractionated treatment but overall treatment time (typically 5–7 weeks) is not long enough to allow the triggering of proliferation in late-responding tissues. Therefore early reactions,
such as reactions of the skin or mucosa, can be easily dealt with by simply prolonging the overall treatment time. However, such a strategy has no effect on the late reactions and, as we will now discuss, may actually have a detrimental effect on the therapeutic efficacy of the treatment if proliferation of tumour cells becomes significant.

Irradiation of tumour cells can trigger the surviving cells to divide faster than before. This is known as accelerated repopulation. This phenomenon has been observed in animal studies and there is evidence for it in some human tumours. Withers et al. (1988) performed a review of tumour control data for head and neck cancer and demonstrated a ‘dog-leg’ variation in total dose for a given tumour control response with treatment time of the form illustrated in Figure R.44. If the treatment is completed before the initiation of accelerated repopulation (denoted by the discontinuity at time $T_{\text{delay}}$), the dose required is fixed and independent of treatment time. However, if the treatment duration is greater than $T_{\text{delay}}$, the dose required to produce the same tumour response increases and is time-dependent. This effect is attributed to tumour repopulation and should be considered in biological effective dose (BED) calculations for schedules where the overall treatment time is extended compared with that prescribed, such as may occur when there is an interruption in treatment. Recent data have identified that prolongation of overall treatment time detrimentally affects local control rates for tumour types including squamous-cell carcinomas (SCC) of the head and neck region, cervix, lung, oesophagus, skin, and possibly vagina. It is also thought to affect medulloblastoma and carcinoma of the bladder. Rapid proliferation does not occur in carcinoma of the breast and prostate and hence overall treatment time is not so critical for these tumour types.

A number of clinical trials have been performed using alternative fractionation schedules to the standard (one fraction per day of around 2 Gy given 5 days a week for up to 7 weeks) to try to reduce the effect of tumour repopulation on local control. Continuous hyperfractionated accelerated radiation therapy (CHART), which reduces overall treatment from 6–7 weeks to 12 days and gives 36 small fractions, has been tested in multicentre randomised controlled clinical trials. The trial in non-small-cell lung cancer showed improvement in survival and this regime is now the government recommended standard of care for eligible patients in the United Kingdom. In the head and neck CHART trial, there was only a small, non-significant improvement in the disease-free interval. However, the DAHANCA Trial 6 and 7 for SCC of the head and neck, initiated to examine whether a reduction of the overall treatment time by increasing the number of weekly radiotherapy fractions from 5 to 6 (while maintaining same total dose and fraction number) improved the tumour response (and were acceptable with regard to early and late morbidity), showed an improvement in 5 year loco-regional control for the 6 fraction per week arm compared with 5 fraction per week arm. This accelerated schedule has become the standard baseline treatment in Denmark.

**Abbreviations:** BED = Biological effective dose, CHART = Continuous hyperfractionated accelerated radiation therapy, DAHANCA = Danish head and neck cancer and SCC = Squamous cell carcinoma.

**Related Articles:** Alpha beta ratio, Biological effective dose, Cell cycle, Cell proliferation, Fractionation, Interruption of treatment, Radiosensitivity, Repair, Redistribution, Reoxygenation, 5Rs of radiobiology, Therapeutic efficacy, Tumour control probability


**Resistance, electrical**

*(General)* Electrical resistance is a ratio of the potential applied to a given conductor to the current intensity value. The SI unit of electrical resistance is the ohm, symbol $\Omega$.

Assuming a uniform current density, electrical resistance of a resistive object is a function of both its physical geometry and the resistivity of the material it is made from:

$$ R = \frac{\ell \cdot \rho}{A} $$

where
- $\ell$ is the length (in m) of the object
- $A$ is the cross sectional area (in m$^2$) of the object
- $\rho$ is the resistivity (in $\Omega$m) of the material

The resistance of a resistive object determines the amount of current through the object for a given potential difference across the object, in accordance with Ohm’s law:

$$ I = \frac{V}{R} $$

where
- $R$ is the resistance of the object (measured in ohms)
- $V$ is the potential difference across the object (measured in volts)
- $I$ is the current through the object (measured in amperes)

The electrical resistivity of a metallic conductor decreases gradually as the temperature is lowered. Near room temperature, electrical resistance of a typical metal increases linearly with increase...
of temperature. Even near absolute zero copper shows a non-zero resistance. In comparison, the electric resistance of a typical intrinsic (non-doped) semiconductor decreases exponentially with increase of temperature.

In certain materials a phenomenon of zero electrical resistance occurs, called superconductivity, found to be present generally at very low temperatures. The resistance of a superconductor drops abruptly to zero when the material is cooled below its critical temperature. The superconductors used in superconducting coils of high magnetic field MRI scanners become superconducting only at temperatures below 10 K and special cooling with liquid helium is needed for this purpose.

**Related Articles:** Insulation resistance, Magnet(s), Supercapacitor, Superconducting magnet, Superconducting material

### Resistance index
(Ultrasound; Clinical; General) The resistance or resistive index is a simple, commonly used measure of arterial waveform shape. It was designed to reflect distal resistance in the cerebral circulation (1). The resistive index (RI) is non-dimensional and independent of the ultrasound beam/blood flow direction angle. The RI is calculated from peak systolic and end diastolic velocities in a flow waveform as shown in Figure R.45 and is calculated by

\[
RI = \left( \frac{V_{\text{max}} - V_{\text{min}}}{V_{\text{max}}} \right)
\]

**Abbreviations:** EDV = End diastolic velocity, PSV = Peak systolic velocity, \( V_{\text{max}} = \) Maximum velocity and \( V_{\text{min}} = \) Minimum velocity.

**Related Articles:** Pulsatility index, Peak systolic velocity, End diastolic velocity

**Further Reading:** Planiol, T. and L. Pourcelot. 1973. Doppler effect study of the carotid circulation. In: Second World Congress on Ultrasonics in Medicine, Rotterdam, the Netherlands, pp. 104–111.

### Resitive magnet
(Magnetic Resonance) In an electromagnet with resistive conductors power loss is lost as coulomb heat and the achievable magnetic field strength is limited. Basic designs for open MRI systems (Figure R.46) are the ‘double donut’ for horizontal \( B_0 \) field and the ‘C-shape’ for vertical \( B_0 \) field. In the latter, the field strength is enhanced by an iron-core. The maximum static magnetic field is 0.6 T for the iron-cored design and 0.2 T for the air-cored design.

**Related Articles:** Electromagnet, Fringe-field, Magnet, Permanent magnet, Superconductive magnets

### Resistor
(General) See Resistance, electrical

### Resolution
(Diagnostic Radiology) Resolution describes the ability to ‘resolve’ or see the separation between two objects.

The ability of an imaging system to resolve two relatively small and close objects is spatial resolution. It is reduced by image blurring. Spatial resolution test objects consisting of adjacent lines separated by spaces (line pairs) are used to evaluate the effect of blurring in imaging procedures. Often the term ‘resolution’ is used as synonym of spatial resolution (measured in line pairs per millimetre, \( Lp/mm \)).

The ability of an imaging system to resolve small differences in contrast (low contrast) of objects is contrast resolution. It is reduced by image noise. Contrast resolution test objects are mainly related to contrast detail measurements.

In digital imaging an image is represented by three main components – matrix height, width and depth (bits, greyscale). The first two are related to the spatial resolution; the depth is related to contrast resolution (noise dependent).

**Related Articles:** Spatial resolution, Contrast resolution, Line pairs, Modulation transfer function, SNR

### Resolution
(Magnetic Resonance) Spatial resolution is a measure of the ability to distinguish closely spaced objects in an image.

Ideally the spatial resolution of MR images should be limited by the size of the voxels in the image matrix. For example, if a 240 mm field of view is imaged using a 256 × 256 acquisition matrix, then the spatial resolution should be 0.9375 mm in both the frequency and phase encoding directions.

In practice, whilst frequency encoding is performed over a single step, phase encoding is performed over a series of repetition times (256 steps for a 256 × 256 matrix), each with a different amplitude. As a result, the phase encoding resolution will be more affected by any inconsistencies in gradient switching rate, gradient linearity and/or eddy current compensation. For this reason the resolution in the phase encoding direction is generally of lesser quality than the resolution in the frequency encoding direction.

**Measuring Spatial Resolution:** Spatial resolution may be assessed visually using images of bar patterns/line pairs. Figure R.47 shows an image of the specially designed EuroSpin Test Object 4.
Here the size of the smallest bar set for which resolution is achieved is taken as the resolution.

**Further Reading:** Ler, R. et al. 1998. Quality control in magnetic resonance imaging, IPEM Report 80, Institute of Physics and Engineering in Medicine, York, UK.

**Resolution** *(Ultrasound)* In imaging, resolution describes the ability of a system to separate objects. In an ultrasound scanner resolution is usually described and measured by

- Spatial resolution, the ability to separate similar objects in the axial, lateral and slice thickness planes
- Contrast resolution, the ability of the display to separate the acoustic properties from different tissues
- Temporal resolution (link), how quickly images are updated

The resolution of ultrasound scanners is dependent on their design and operation. For example, poor control of beam width leads to a reduced lateral resolution. Scanning deep structures may lead to a reduction in temporal resolution. The performance of ultrasound scanners is limited by the pulse-echo time required \((1/PRF)\) and the number of pulses required to form images. The best use of this time is one of the main considerations in scanner design; users are often faced with choices to optimise temporal resolution or spatial resolution of an image.

**Resonant frequency** *(Magnetic Resonance)* Classically, under the influence of a static external magnetic field \(B_0\), a magnetic dipole with an orthogonal component \(\gamma h\) precesses with angular frequency \(\omega_0\):

\[
\omega_0 = \gamma B_0
\]

where \(\gamma\) is the gyromagnetic ratio for a specific nucleus. Hence, the resonant frequency \(\omega_0\) is commonly known as the Larmor frequency (see Larmor frequency article).

Quantum mechanically, the energy difference between the two levels of a spin-1/2 nucleus in an external magnetic field (Zeeman splitting, Figure R.48) may be associated with a resonant frequency according to: \(E = h\nu\) This article will focus upon resonance effects from a quantum mechanical point of view; for a classical approach see the RF pulse article.

The energy difference between the two levels on Figure R.48 has an associated frequency

\[
\nu = \frac{\gamma}{2\pi} B_0
\]

which is the resonant frequency.

**Resonant Frequency Effects:** Transitions between the two spin states shown in Figure R.48 are electric dipole forbidden but can be induced by magnetic fields. This can be achieved by applying a strong static field, but is more easily achieved using a field oscillating at the resonant frequency of the system, allowing a much lower strength field to be used.

Quantum mechanically, a spin with \(N\) valid energy states can also exist in any linear superposition of these \(N\) states. Oscillating magnetic fields at resonance with the spin drive it back and forth between the energy levels, passing through all the superposition states along the way. RF pulses are described by quantum statistics of a spin ensemble. Thus, by choosing the duration of the RF pulse it is possible to generate any desired superposition.

In MRI the classically termed 180$^\circ$ pulse corresponds quantum mechanically to a direct interconversion between the two states shown in Figure R.48. The classical 90$^\circ$ pulse causes an equally weighted superposition of the two spin states in the quantum mechanical picture.

![Figure R.47](image1.png) **Image of EuroSpin Resolution Test Object (TO4).**

![Figure R.48](image2.png) **The energy of a spin-1/2 magnetic moment \(\mu\) split by a \(B\)-field as seen in quantum mechanics.**
Quantifying the Resonant Frequency: From Figure R.48, the frequency of the transition between the energy levels of a spin-1/2 nucleus is given by $v = (\gamma/2\pi)B_0$

For a proton, $\gamma = 2.68 \times 10^8 \text{rad/s/T}$. Therefore, at a field strength of 1.5 T, the resonant frequency of the proton is

$$v = \frac{2.68 \times 10^8 \text{rad/s/T}}{2\pi} \times 1.5 \text{T}$$

$$v = 63.98 \text{ MHz}$$

In practice if an MR scanner is used off-resonance, it will produce images with reduced SNR; thus, the resonance should be checked prior to clinical acquisition by QA measurements.

Related Articles: RF pulse, Larmor frequency

Respiratory gating

(Radiotherapy) Respiratory gating is important in two areas: three-dimensional imaging of mobile anatomy and treatment of mobile anatomy using external beams.

Image Acquisition: Tomographic imaging of mobile organs may be subject to geometric distortion due to interplay between the scanning time and the breathing or cardiac cycle. A consequence of this is that a spherical object may appear distorted or even as two or more objects in a multislice CT scan (Reitzel et al.). Respiratory gating is used to ameliorate these effects. Two approaches have been used:

1. To acquire the scan while the patient is holding their breath.
   This is a simple solution but has the disadvantage for radiotherapy treatment that the technique cannot be used during treatment (owing to the long beam-on time) and thus is not likely to be representative of the treatment-time position.
2. To acquire the raw scan data with a measurement of position in the breathing or cardiac phase.

In this case the data may be used to form an image of a particular phase, or to create a time sequence of images throughout the breathing or cardiac process, in so-called four-dimensional CT (4DCT). In the case of the cardiac cycle, an electrocardiograph (ECG) is generally used. To measure the breathing cycle, markers on the patient surface, spirometry, or measurement of the temperature of air at the patient’s nostrils have been used.

Radiotherapy Treatment: In radiotherapy of treatment sites with organs that are mobile due to respiration (including lung and liver), it is desirable to control the effects of this motion. This has the benefit of reducing the risk of geometric miss and allowing margin reduction, with consequent sparing of normal tissues. Two basic approaches to gating are used:

1. To gate the treatment delivery.
   This involves measuring the motion and turning the treatment beam on and off at the appropriate phase of the motion. Several signals may be used to control this process. External markers placed on the patient surface and x-ray imaging of implanted markers are two examples.
2. To gate the patient’s breathing.
   This involves either using a device to hold the patient’s breath at a particular phase of the breathing cycle (usually inhale or exhale) or asking the patient to hold their breath at inhale or exhale in a voluntary breath-hold procedure. The first of these approaches involves using a spirometer to measure the patient’s breathing and a valve to hold the breath at the desired phase. Active breathing control (ABC) is the system that implements this. The beam-on time of a radiotherapy treatment is generally longer than a patient can comfortably perform a single breath hold and thus the treatment is often delivered in several breath holds with a pause in treatment in between.

Abbreviations: ABC = Active breathing control, CT = Computed tomography, ECG = Electrocardiograph and 4DCT = Four-dimensional computed tomography.

Related Articles: Radiotherapy, Intrafraction movement, Margins, Deep inspiration breath hold, Gating – respiratory


Restricted area

(Radiation Protection) Restricted area is a general term used to describe an area or room where access to the area must be controlled to avoid persons being harmed. Only persons authorised to do so may enter a restricted area, either following written procedures, or wearing personal protective equipment (PPE).

In the United Kingdom, the Ionising Radiation Regulations (IRR99) describes a two-tier system of designation given to restricted areas based on the level of radiation hazard and risk present:

Controlled Area: ‘Any area in which it is necessary to follow special procedures to limit exposure, or any area where exposure to > 3/10 of any radiation dose limit is possible’.

Supervised Area: ‘Any area in which it is necessary to review the need for future control, or any area where it is possible for an individual to receive an effective dose of 1 mSv y\(^{-1}\). In other words an area should be supervised if a worker is liable to receive more than the public dose limit.

All restricted areas must be appropriately demarcated. Ordinarily inside a building, a controlled or supervised area will be defined by the walls of a room which act to shield persons outside the room from the hazard. However, it may be necessary to define an area either within a room, or outside the room, or indeed the building as a restricted area, and to demarcate the area to warn persons of its existence.

Restricted areas (controlled or supervised) must have warning signs to describe the hazard and risk, and to state that only authorised persons are allowed to enter.

Related Articles: Controlled area, Supervised area

Restricted collisional mass stopping power

(Radiation Protection) The restricted collisional mass stopping power describes how charged particle ionising radiation is absorbed in collisions with atomic nuclei in an absorbing medium, but only those interactions that lead to the total energy of the incident particle being absorbed either at the site of the interaction, or very close by, such that there is no radiative loss out. This concept is the most useful at the microscopic level in describing how energy is deposited and leads to absorbed dose.

For further information, see the articles on Collisional mass stopping power and Absorbed dose.

Related Articles: Collisional mass stopping power, Absorbed dose

Restricted stopping power

(Radiotherapy) The stopping power indicates the average rate of energy loss by a charged particle in a medium and this is not always equal to the absorbed dose to the medium particularly if the range of the secondary electrons produced is large. In fact a secondary electron with enough energy, called delta ray, can leave the immediate vicinity of the primary particle path and produce its own track. The concept of restricted stopping power has been introduced to
associate an energy loss in a medium more closely to absorbed dose to the medium. The restricted stopping power is defined as the linear rate of energy loss due only to collisions in which the energy transfer does not exceed a specific value $\Delta$. The extent of the localization is determined by the size of the cut-off energy $\Delta$.

**Retention fraction**  
*(Nuclear Medicine)* Retention fraction is the fraction of a radio-pharmaceutical delivered to an organ or tissue that is extracted into and retained by the tissue. This fraction is the residual after clearance of the vascular component and the portion of radiopharmaceutical that rapidly diffuses back from the organ or tissue into blood.

**Retina**  
*(General)* The retina forms the internal, light-sensitive, layer of the eye, onto which light is focused by the cornea and lens. The electrical signals generated by the interaction of light with the retina’s photoreceptor cells are transmitted to the brain via the optic nerve. The point at which this connects to the retina is commonly known as the ‘blind-spot’. The retina is approximately 42 mm in diameter, and 0.5 mm thick. The greatest resolution is achieved when light is focused onto fovea, the central portion of the retina.

The retina contains two types of photoreceptor cells, rods and cones. There are around 125 million rods cells, each having a diameter of 0.002 mm. They are found mostly in the peripheral retina and are more sensitive to low-level light and hence are responsible for peripheral and night vision. There are approximately 6 million cones with a diameter of 0.006 mm, the majority of which are found in the fovea. They are less sensitive and hence are responsible for vision in bright light and also colour vision.


**Reverberation**  
*(Ultrasound)* Reverberations are artefacts caused by re-reflection of reflected ultrasound echoes. They occur most commonly in the near field if strong reflecting interfaces are present normal to the beam direction, parallel to the beam.

Re-reflection often occurs at the tissue-transducer interface due to the mismatch of acoustic impedance at this surface (Figure R.49). Reverberation artefacts are most obvious if there are weak echoes from direct imaging of the tissue, for example in veins, cysts and amniotic fluid (Figure R.50) but they can also obscure detail in greyscale imaging (Figure R.51) by superimposing extraneous echoes to those arising from the target tissue. Closely spaced reverberations from small structures can be described as comet-tail artefacts.

**Related Articles:** Acoustic impedance, Reflection, Comet tail

**Reynolds number**  
*(Ultrasound; General)* The Reynolds number describes ratio of inertial forces to viscous forces in fluid flow and so indicates their relative dominance in given flow conditions.

For flow in a tube, Reynolds number is defined as

$$\text{Re} = \frac{\rho V D}{\mu}$$

where
- $\rho$ is the density of the fluid
- $V$ is the velocity
- $D$ is the diameter of the tube
- $\mu$ is the fluid viscosity
The number is dimensionless and the higher the number the greater the tendency towards turbulence. In a tube with steady laminar flow the transition to turbulent flow occurs at a Reynolds number of about 2300 although the precise value is dependent on local conditions. Around this value, flow is transitional, neither laminar nor turbulent. Flows with Reynolds numbers <2000 are laminar and >3000 are turbulent. In the case of pulsatile flow the peak velocity may be used and the number is described as the peak Reynolds number.

**Related Articles:** Laminar flow, Turbulent flow

**RF (radiofrequency)**

(Magnetic Resonance) See Radiofrequency (RF)

**RF coil**

(Magnetic Resonance) An RF coil in MRI receives and/or transmits radiofrequency energy into the patient. A standard MRI system will have many RF coils, each appropriate to a specific range of imaging tasks.

An RF coil forms a tuned circuit with a resonant peak. Coils are designed to resonate at the Larmor frequency for the MRI field strength in question, that is for a 1.5 T MRI, coils will have a resonant peak for protons at approximately 64 MHz and lower for other nuclei.

For a receiver coil it is desirable that the coil displays high SNR, good uniformity and adequate coverage. Good uniformity equates to a uniform gain or sensitivity to signal throughout a tissue volume. A volume coil is designed to enclose the anatomy of interest and provide high uniformity. A surface coil is designed to be placed in proximity to more superficial anatomy (e.g. the spine). A surface coil provides better SNR for the anatomy image but a reduced field of view and reduced uniformity.

In a transmit coil, the coil generates a magnetic field at right angles to the main static $B_0$ field when the RF system applies a pulse to the coil. This is called the $B_1$ field. A component of this magnetic field rotates in the same sense as the direction of precession and causes spins to flip though an angle. In order to generate uniform flip angles throughout a volume, the energy deposited by a coil should be uniform throughout the tissue volume.

In coil design, coil transmission and reception characteristics are linked by the principle of reciprocity. If a coil demonstrates good uniformity of sensitivity while acting as a receiver, the same coil will also demonstrate good uniformity in generating flip angles if acting as a transmitter.

Because surface coils demonstrate poor uniformity, they are not normally used as transmit coils. The body coil is generally used as the transmit coil for surface coils and other volume coils. The body coil encloses the cylindrical bore internally in the scanner and surrounds the patient. A coil that acts as both a transmitter and receiver is called a transceiver.

An RF coil could be as simple as a loop of wire with a tuning capacitance added to set the appropriate resonant peak. In practice coil designs adopt a range of conductor geometries in order to maximise SNR or uniformity for a given application. Standard geometries used in practice include coils based on solenoid, birdcage and saddle designs.

**RF echo**

(Magnetic Resonance) A radiofrequency (RF) echo is a rarely used term for ‘spin echo’ to distinguish it from a gradient echo. A few authors use RF echo for a spin echo where the flip angle of the refocusing pulse is less than 180° and greater than 90°.

**Related Articles:** Radiofrequency, Spin echo


**RF pulse**

(Magnetic Resonance) A radiofrequency pulse is a controlled emission of electromagnetic radiation in the radio frequency part of the spectrum. It is an important feature of MRI because the resonant frequencies of the nuclei being used lie in this part of the spectrum.

This article describes basic effects of ‘hard’ (rectangular) RF pulses from a classical point of view: for a quantum mechanical explanation see the Resonant frequency article.

Classically, under the influence of an external magnetic field $B_0$, nuclear spins precess with the Larmor frequency $\omega_0$:

$$\omega_0 = \gamma B_0$$

where $\gamma$ is the gyromagnetic ratio for a specific nucleus.

In NMR only the magnetic component $B_1$ of the RF pulse is considered in the rotating frame. When $B_1$ is applied on the resonant frequency and perpendicular to $B_0$, the net magnetisation $M$ will experience a torque which causes it to rotate towards the transverse plane.

The angular frequency of nutation by the additional field $B_1$ is

$$\omega_1 = \gamma B_1$$

from which the following expression can be obtained:

$$t = \frac{\theta}{\gamma B_1}$$

which relates the duration of the RF pulse ($t$), to the angle through which $M$ is rotated ($\theta$ in rads).

For maximum MR signal, the transverse component of $M$, is maximised, such that it is rotated through 90° from the z-axis into the x-y plane. Using the equation above the exact duration of the RF pulse to achieve this rotation can be calculated by setting $\theta = \pi/2$. Such a pulse is known as a 90° RF pulse.

Another key pulse is the 180° pulse which is either used to invert longitudinal magnetisation, or for refocusing transverse magnetisation to correct for dephasing caused by the imperfect $B_0$ field.

**Related Article:** Resonant frequency

**RF uniformity**

(Magnetic Resonance) Radiofrequency uniformity can be divided into RF transmission uniformity and RF reception uniformity. The RF uniformity is governed by which RF coil is used. The best RF homogeneity is obtained by using the most appropriate coil and ensuring that the coil is positioned correctly with respect to the patient and perpendicular to the static magnetic field.

Uniformity is important, especially for transmit and receive coils as poor uniformity will affect RF flip angles and therefore image contrast. Large transmission coils give a very uniform transmission field but are not very sensitive for receiving the signal. Receive only coils are often non-uniform in design to obtain the best SNR. Surface coils are the most prone to non-uniformity, providing high SNR close to the surface at the expense of non-uniformity. Surface coils are thus typically used to image anatomical structures close to the surface of the patient.
To measure the effects of RF uniformity, a uniform phantom is imaged. The phantom should be loaded to simulate the presence of a patient. An ROI covering 75% of the slice is then analysed. Integral uniformity equation is given by

\[ I = 1 - \frac{(M + m)}{(M - m)} \times 100\% \]  

(R.2)

where \( M \) is the maximum pixel value and \( m \) is the minimum pixel value within ROI of 75% of the phantom area. 100% is perfect uniformity.

**Abbreviations:** RF = Radiofrequency, ROI = Region of interest and SNR = Signal-to-noise ratio.

**Related Articles:** B, homogeneity, B, inhomogeneity

**R**, value
*(Nuclear Medicine)* R, value is the retention factor, a value used in thin-layer chromatography for identification of components in radiopharmaceuticals. The R, value is defined as

\[ R_f = \frac{\text{migration distance of the component}}{\text{migration distance of the solvent}} \]

The values range from 0 to 1 and are characteristic for a given compound provided that the same stationary and mobile phases are used.


**Rhenium** *(General)*

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Re</th>
</tr>
</thead>
<tbody>
<tr>
<td>Element category</td>
<td>Transition metal</td>
</tr>
<tr>
<td>Mass number A</td>
<td>186</td>
</tr>
<tr>
<td>Atomic number Z</td>
<td>75</td>
</tr>
<tr>
<td>Atomic weight</td>
<td>186.207 kg/kg-atom</td>
</tr>
<tr>
<td>Electronic configuration</td>
<td>1s² 2s² 2p⁶ 3s² 3p⁶ 3d¹⁰ 4s² 4p⁶ 4d¹⁰ 4f¹⁴ 5s² 5p⁶ 5d¹ 6s²</td>
</tr>
<tr>
<td>Melting point</td>
<td>3459 K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>5869 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>20,800 kg/m³</td>
</tr>
</tbody>
</table>

**History:** Rhenium is a high-density, high-melting-point silvery grey transition metal. It was first discovered in 1925 by spectroscopic analysis of certain mineral ores in which it is present in trace amounts. In 1928 the discoverers succeeded in extracting 1 g of the metal from 660 kg of molybdenite ore. Rhenium is still obtained only as a by-product of other metal refining processes and is one of the world’s most expensive metals.

**Medical Applications:** X-ray tube target – In a rotating anode x-ray tube, the focal track for electrons is usually made of a mixture of tungsten and rhenium. Both of these materials have a high atomic number, a high specific heat capacity and a high melting point. However, as tungsten has low linear expansivity it is prone to crazing with repeated expansion and contraction. Rhenium has a higher linear expansivity and thus slows the rate at which crazing occurs.

**Radionuclide therapy – In nuclear medicine therapy rhenium is used in two isotope forms:** ¹⁸⁶Re and ¹⁸⁸Re.

<table>
<thead>
<tr>
<th>Isotope of rhenium</th>
<th>¹⁸⁶Re</th>
</tr>
</thead>
<tbody>
<tr>
<td>Half-life</td>
<td>89h</td>
</tr>
<tr>
<td>Maximum decay energy, ( E_{\text{max}} )</td>
<td>( \beta: 1.07 \text{ MeV} ), ( \gamma: 137 \text{ keV} )</td>
</tr>
<tr>
<td>Isotope of rhenium</td>
<td>¹⁸⁸Re</td>
</tr>
<tr>
<td>Half-life</td>
<td>17h</td>
</tr>
<tr>
<td>Maximum decay energy, ( E_{\text{max}} )</td>
<td>( \beta: 2.10 \text{ MeV} ), ( \gamma: 155 \text{ keV} )</td>
</tr>
</tbody>
</table>

The beta particles emitted from ¹⁸⁶Re have a range of 5 mm in water whilst those emitted from ¹⁸⁸Re have a range of 11 mm, thus ¹⁸⁶Re is typically used for small tumours whilst ¹⁸⁸Re is more appropriate for larger masses.

While ¹⁸⁶Re is generated via neutron radiation of ¹⁸⁵W, although it is not as common as ⁸⁹Sr-Metastron or ¹⁵³Sm-EDTMP. ¹⁸⁶Re emits a beta particle with maximum energy 1.07 MeV, mean energy 0.349 MeV, average soft-tissue range 1.1 mm and a gamma photon of energy 137 keV (9%). The physical half-life is 3.72 days. Whereas the beta radiation is used to irradiate the cancer, the emitted gamma radiation can be used simultaneously for imaging and quantification of the ¹⁸⁶Re uptake.

The absorbed dose to bone surfaces and red bone marrow are 1.4 and 1.3 mGy/MBq, respectively. Other organs with significant absorbed doses are the walls of LLI and urinary bladder, 0.57 and 0.54 mGy/MBq respectively. Most other organs receive about 0.02 mGy/MBq. The effective dose for ¹⁸⁶Re-HEDP is approximately 0.3–0.4 mSv/MBq.


**Related Articles:** Tc-⁹⁹m-diphosphonates, Sm-153-EDTMP [Lexidrom], Sr-89-chloride [Metastron]

**Rheostat** *(General)* See *Resistance, electrical*

**Rigid stem chamber** *(Radiation Protection)* The rigid stem chamber is a cylindrical ionization chamber to be used for absolute dose measurement in radiotherapy with high-energy photon and electron beams.
The chamber is not waterproof and can be used in solid state phantoms for absorbed dose to water, air kerma and exposure measurements.

Related Article: Cylindrical ionisation chamber

Further Reading: http://www.ptw.de

Ring artefact

(Diagnostic Radiology) Ring or band artefacts in a CT image are a result of an incorrect signal from a detector channel or a group of detector channels (Figure R.52). This could be due to faulty detectors, a problem in the data acquisition system electronics, or incorrect calibration. The artefact is mainly seen on third generation CT scanners.

Figure R.53 illustrates the origin of these artefacts. The path of an x-ray from the source to a particular detector is at a fixed distance from the isocentre for all angular positions. A defective detector will therefore form a ring pattern during the reconstruction process.

Related Articles: Artefact, Beam hardening, Computed tomography, Cone beam artefact, Helical artefact, Image artefact, Metal artefact, Motion artefact, Partial volume effect (artefact)

Ring artefact

(Magnetic Resonance) The ring artefact is similar to Gibb’s artefact. Bands of high- and low-intensity pixels form parallel to a high contrast edge due to undersampling of the data in the phase-encoding direction.

The high contrast edge in question is caused by the introduction of gadolinium into the arteries during contrast enhanced MRA. If the centre of k-space is being sampled when the signal intensity changes rapidly, the line in k-space will have a different contrast weighting, causing the artefact to occur.

Abbreviation: MRA = Magnetic resonance angiography.

Related Article: Gibb’s artefact

Ring artefact

(Nuclear Medicine) A common image artefact. As the name indicates the phenomenon is one or a series of rings in the image caused by a systematic error. The source of error differs between camera systems and some image reconstruction methods can cause ring artefacts. For example, using window filters with sharp edges when reconstructing can lead to ring artefacts (also called Gibbs phenomena).


Ring down

(Ultrasound) Ring down is a reverberation artefact where a pair of closely spaced reflectors, most commonly the edges of a gas bubble, cause a continuous echo in the image from the reflectors in the direction of the beam.

Related Articles: Comet tail, Reverberation

Ripple

(Diagnostic Radiology) See Voltage waveform

Ripple factor

(Diagnostic Radiology) The ripple factor is maximal change (drop) of kV, compared with the maximal kV (kVp). The ripple factor is expressed as a percentage of the kV_{min} from kV_{max} (kVp). The factor is directly related with the assessment of the kV waveform of an x-ray generator. For more information see the article on Voltage waveform. From there one can see that a two pulse kV waveform
Rise time

(Magnetic Resonance) When a magnetic field gradient is switched on during an MRI pulse sequence, it takes a finite length of time for the current passing through the gradient coils to rise from zero and hence for the gradient field itself to reach the required value. This period of time is known as the rise time, and is dependent on the resistance and inductance of the gradient coils (Figure R.54).

Shortening the rise time allows for shorter gradient pulses, which increases imaging speed and can be advantageous in other ways for specific MRI applications. Thus the minimum rise time to maximum gradient amplitude is an important system performance parameter.

The related parameter of gradient slew rate is often defined as the maximum gradient amplitude divided by the minimum rise time. However, the maximum slew rate that can be delivered by a system may not be sustainable over the rise time needed to reach maximum amplitude, in which case this definition may be misleading.

Rise times are typically on the order of a few hundred microseconds on modern clinical MRI systems.

Risk assessment

(Radiation Protection) Risk is defined as the chance, probability or likelihood that a person may experience an adverse effect from exposure to a hazard. Risk can be quantified as a risk coefficient in several ways – for example the risk may be quoted as 1 in 10, 10%, 0.1 or $1 \times 10^{-1}$.

A hazard, or health hazard, is defined as anything that has the capacity to cause harm to a human being. Such harm can be through physical injury, biological damage or other health detriment (adverse effect) in the form of cancer or other disease. Examples of hazards include fire, chemicals, electricity (electrocution), water (drowning), and exposure to ionising or non-ionising radiation.

Risk assessment is generally accepted to be a five-stage process:

- **Identify the hazard** – what is the source of the hazard (e.g. fire, ionising radiation exposure, etc.).
- **Identify those who might be harmed** – there will be ‘critical groups’ of individuals who may be most at risk of suffering harm from exposure to the hazard. It is therefore most important to consider reducing the exposure of these individuals first.
- **Quantify the risk** – what are the chances that the critical groups of individuals identified earlier may come to harm.
- **Identify control measures** – decide on an action plan to reduce the risks of harm to acceptable levels.
- **Review the efficacy of actions taken** – analyse the effect of the control measures and revise the action plan if necessary.

For example, in radiation protection, control measures can be considered as a hierarchy of systems in place to reduce radiation dose to individuals. One can consider the boundaries of a room containing the radiation hazard, the walls, ceiling, floor, and the doors. Shielding the boundaries or through restriction of access and the use of interlocks to prevent a person from being exposed to ionising radiation may be an ideal solution to reduce radiation dose.

If however it is necessary for a person to be in the room during the exposure to carry out specific procedures, then it will be necessary to establish a system of work for the procedures. Finally, if the system of work is not able to reduce doses to acceptable levels, then personal protective equipment must be used (lead aprons, etc.).

**Related Articles**: Absolute risk, Adverse effect, Hazard, Radiation risk, Risk coefficient

Risk coefficient

(Radiation Protection) Risk coefficient is the quantification of the chance or probability that a hazard may occur, for example the risk may be quoted as 1 in 10, 10%, 0.1 or $1 \times 10^{-1}$.

**Related Article**: Risk assessment

Road mapping

(Diagnostic Radiology) Road mapping (RM) can be seen as advanced last image hold (LIH). RM is used in angiography with digital fluoroscopy systems. Initially an image of the vessels is made by injecting small amount of contrast media in the vessels where the catheter will be introduced. This ‘vessels map’ is memorized (as LIH) and is later superimposed over the live fluoroscopy during the introduction of the catheter. This way the radiologist/angiologist has good guidance through the patient vessels anatomy. The RM method is primarily based on software.

**Related Articles**: Last image hold, Digital fluoroscopy

Roam and zoom

(Diagnostic Radiology) High-resolution digital images (as digital mammography) require special display screens. However even the best contemporary monitors are not large enough to display at once a full digital mammogram with its best resolution. Due to this reason the image has to be first zoomed (to get the best resolution) and after this displayed partially through the roam function of the equipment.


Robotic linac

(Radiotherapy Equipment) Various technologies have been developed to improve both the dose distribution and positional accuracy of radiotherapy. The robotic linac is a concept which attempts to achieve both goals. Such systems are often used for radiosurgery, in which the treatment is delivered in a single fraction. Examples of robotic linacs include the Accuray Cyberknife.

**Robotic Delivery System**: The delivery system consists of an industrial robot with a small, x-band linear accelerator.
The accelerator may be orientated at a variety of angles in three dimensions to enable the target to be irradiated over a large solid angle (typically 2.5π steradians) with the total dose delivered in many beamlets with small dose irradiations over numerous positions.

**Verification System:** The position of the patient’s anatomy for each beamlet is often verified with an imaging system. This is usually a combination of projection x-rays and fiducial markers for each beam position coupled to more continuous imaging of the external anatomy using markers placed on the patient’s skin.

**Related Articles:** Image-guided radiotherapy, Intensity modulated radiotherapy, Radiosurgery


**ROC (receiver operating characteristic)**

(Diagnostic Radiology) See Receiver operating characteristic (ROC)

**Röentgen (R)**

(General) This was the first quantitative measure of radiation initially defined by the ICRU (International Commission on Radiation Units) in 1928 as the quantity of x-rays when the secondary electrons are fully absorbed and any wall effects avoided would produce 1 esu of charge in 1 cc of air at 0°C and 76 cm mercury of pressure.

The unit was later redefined (1937) as the quantity of x or gamma radiation producing 1 esu of charge per 0.001293 g air. (The mass of air is that which is associated with 1 cc of air.)

This unit was subsequently replaced by the SI unit of exposure in C/kg:

1R = 8.69 mGy
1R = 0.258 mC/kg

The unit is named after Wilhelm Röntgen German physicist, discoverer of x-rays (1845–1923).

**Related Article:** Exposure

**ROI (region of interest)**

(Nuclear Medicine) See Region of interest (ROI)

**Roos chamber**

(Radiation Protection) The Roos chamber is a waterproof plane parallel ionisation chamber used for absolute dosimetry in high-energy electron and proton beams as well as for measuring photon depth dose. This chamber can be used in water or in solid state phantoms for absorbed dose to water measurements.

**Related Articles:** Absorbed dose, Parallel plate ionization chamber

Further Reading: [http://www.ptw.de](http://www.ptw.de)

**Root mean square (RMS) voltage**

(General) The voltage value that has the same effect and gives the correct result on a power calculation as does a DC voltage of the same value. RMS voltage is equal to the square root of the mean value of the squares of the magnitudes of an AC voltage measured at each instant over a defined period of time, usually one cycle.

The RMS voltage is also known as the effective voltage (see the eponymous article).

**Related Article:** Effective voltage value

**Rose model**

(Diagnostic Radiology) The Rose model is a classical model describing image quality based upon quantum noise and its affect on human perception. It is used in medical imaging as a simplistic model to describe the quantum noise of a diagnostic image and how it affects the detection of a signal. The model is of limited use in its original form as being only valid for low contrast, uncorrelated Poisson distributed noisy backgrounds and simple objects. However, this model forms the basis for more advanced image metrics based upon Fourier analysis such as noise power spectrum (NPS) noise equivalent quanta (NEQ) and detective quantum efficiency (DEQ).

The model was first proposed by Albert Rose in 1948, and describes the minimum SNR needed to visually detect a uniform circular object in a uniform background. Historically, Rose’s model was important as it was the first to evaluate the performance of imaging devices by quantum efficiency using an absolute scale.

Rose investigated the detection, by human observers, of circular objects of differing sizes and contrast within a Poisson distributed noisy background. The observers were required to indicate whether the objects were visible or not, while already knowing the presence of the object. This is similar to the Leeds test phantom used in diagnostic radiology quality control today, see Figure R.55.

The Rose model assumes an uncorrelated Poisson distributed noise throughout. If an image had a uniform background with mean quanta per area $q_R$, and an object with mean quanta per area of $q_O$, then the contrast is defined as

$$ C = \frac{q_O - q_R}{q_R} $$

The signal $S$ of the object is then defined as the difference between the mean quanta of the background and the mean quanta of the object, integrated over the area of the object, such as

$$ S = A \left( q_O - q_R \right) $$

The noise in the signal is defined as the standard deviation of the number of quanta in a section of uniform background of area $A$, $\sigma_q$.

For an uncorrelated background this noise is described by Poisson statistics so that

$$ \sigma_q = \sqrt{A q_R} $$

**FIGURE R.55** Leeds test phantom. (Graphs courtesy of EMERALD project, www.emerald2.eu)
Thus, the SNR is given by

$$\text{SNR} = \frac{A(q_B - q_0)}{\sqrt{Aq_B}} = C \sqrt{Aq_B}$$

Rose deduced that a SNR of 5 or above is needed to detect an object under these conditions. In similar circumstances, such as that using the test phantom, with low contrast and uncorrelated Poisson distributed noise, this relationship has been shown to be correct.

Before this model could be used clinically all limitations must be considered. For, one of the main practical limitations of the model is the definition of noise, as clinical images rarely have Poisson distributed noise. For example, electronic noise is usually additive and secondary quanta such as that produced by a phosphor screen shows statistical correlations. Also, anatomical noise can inhibit the detection of signal (pathology) within an image and this has been shown to increase with contrast, whereas the detection of the Rose model object should decrease.

**Related Articles:** Anatomical noise, Detective quantum efficiency (DQE), Noise equivalent quanta (NEQ), Noise power spectrum (NPS), Signal to noise ratio (SNR)


### Rotating anode

*(Diagnostic Radiology)* The first x-ray tube with rotating anode has been produced by Philips in 1929. The main idea of this tube is that the electrons hit different parts of the rotating target, thus increasing many times the size of the thermal focus (distributing the heat in a long target track). At present these are the most-often-used x-ray tubes (with many different designs). Figures R.56 and R.59 show schematically a rotating anode, with indication of its main parts and the three main foci of a rotating anode.

The maximal permissible power of the rotating anode \(P_{\text{max}}\) depends on the effective focal spot size \(f\), the diameter of the target track \(D\), the angle of the anode \(\alpha\) and the speed of rotation \(n = \text{rpm}\):

$$P_{\text{max}} \sim \frac{f^{3/2} D^{1/2} n^{1/2}}{\sin \alpha}$$

![FIGURE R.56 Rotating anode: (1) Real focus (momentary focus); (2) thermal focus (thermal path or thermal track); (3) effective (optical) focal spot.](image)

The rotating anode is mounted on a stem, which is attached to the rotor of an electrical motor (see the article Anode). The stem and the base of the rotating anode are made from tungsten–molybdenum alloy, which is light and decreases the undesirable inertia of the spinning anode. Thin tungsten–rhenium alloy (1–2 mm thick) is coated over the molybdenum alloy, thus forming the rotating target (the target track). In these x-ray tubes the rotating anode and the rotor are within the glass envelope (supported by ball bearings), and the stator producing the rotating magnetic field is fitted outside the glass envelope (Figure R.57). The diameter of the rotating disk varies but often is 100–130 mm.

Increasing the speed of rotation (most often tripling the normal ~3000–9000 rpm) leads to increase of the tube power because the bombarded area of the track passes quickly over the fixed electron beam area and has time to cool during the remaining rotation. Further demand for long sequences of powerful exposures (especially in angiography and CT) lead to constructing rotating anodes with bigger heat storage capabilities. For this purpose the rotating disk is often constructed by molybdenum with embedded graphite compound, which is used as thermal accumulator, thus increasing the heat absorbing volume. This compound has the advantage to be cheaper and lighter than the solid molybdenum, thus diminishing the inertia and decreasing the load on the anode ball bearings (Figure R.58). The thickness of the molybdenum base of the anode disk varies normally from 5 to 15 mm, and the graphite compound could add similar thickness to the anode disk base. This way the overall weight of the anode of a powerful x-ray tube could reach 2 kg.

Although most of the rotating anodes use the basic design described earlier, there are many different designs of diagnostic x-ray tubes. Some improvements of the design of x-ray tubes include use of titanium, zirconium or molybdenum (in mammography) as target, special technologies for building of the target, etc.
Other rotating anodes have radial slits on the tungsten surface that diminishes the effect of thermal expansion and contraction of the metal (thermal stress). Figure R.59 shows only four such slits, while usually there are eight slits or more.

Usually rotating anode x-ray tubes have two focal spots (double focus, or dual filament tubes) and in this case special care is taken for the positioning of the two cathode filament wires. Figure R.59 presents the case when the two filaments overlap, this way creating an area of the anode target, which is heated by both foci. This area is most likely to crack quickly (due to thermal stress), what will decrease the life of the tube. This could be avoided by different positioning of the filament coils.

**Related Articles:** Anode, Stationary anode, Target, X-ray tube housing, Anode angle, Dual filament tube, Bearing

**Rotating frame**

(Magnetic Resonance) It is often helpful to consider the nucleus in a different frame from the laboratory frame, especially to understand the action of RF pulses.

Considering the main magnetic field, \( B_0 \) which is aligned along the \( z \)-axis, and causes spins to precess at the Larmor frequency:

\[
\omega_0 = \gamma B_0
\]

If a frame is chosen which rotates at the same angular frequency as the spin (\( \omega_0 \)), then the spin appears not to precess, and thus according to (R.2) the axial magnetic field in this frame must be zero. This frame is known as the rotating frame, and by choosing it, the main, static magnetic field is effectively cancelled out. Its axes are usually marked by a prime. The rotating frame corresponds to the interaction picture in quantum statistics (Figure R.60).

The second field to be considered is the magnetic component of the radio frequency control pulse, which is a linearly polarised field in the plane transverse to the static field. A linearly polarised field may be considered as a pair of counter rotating circularly polarised fields, and it is easier to apply the frame transformation to these components separately rather than both together. In the rotating frame, the result is two magnetic fields rotating at

![Figure R.58](a) X-ray tube with rotational anode with graphite compound below the molybdenum body (a), compared with the same anode without graphite (b), but with less thermal capacity.

![Figure R.59](a) Rotating anode with its main parts and focal spots. Note the superimposed actual focal spots from both filaments (large LF and small SF), which create two effective focal spots – broad focus (BF) and fine focus (FF).
Rotational 3-D scanning

RF Field 1, rotating with frequency: \( \omega_0 + \nu \)
RF Field 2, rotating with frequency: \( \omega_0 - \nu \)

If an ensemble of spin is exposed to a magnetic field where its rotation period is short compared to the application time of the field, the time-averaged effect of the field will be almost zero because any rotation due to the field will be cancelled out by a subsequent rotation when the field is in the opposite direction a short time later. Thus the sum frequency component (RF field 1) can be safely ignored.

A frequency offset, that is a residual precession in the rotating frame, can be described by a hypothetical z-component yielding the effective \( B_z \):

\[
B_z (\nu - \omega_0) = (B_{1z} (\nu - \omega_0) B_z (\nu - \omega_0), (\nu - \omega_0)/\gamma)
\]

Rotational 3-D scanning

(Nuclear Medicine) Rotational 3-D scanning is another way of describing single photon emission computed tomography or SPECT. A gamma camera system consisting of one or more heads is rotated around the patient, and a number of images are obtained at equal angular intervals over 360° or 180° (depending on the area of the body being scanned). From these images a series of count profiles for each transaxial slice are extracted and reconstructed using filtered back projection or iterative reconstruction. These transaxial slices can then be combined using computer software to provide a three-dimensional image.

Related Article: Single photon emission computed tomography (SPECT).

Rotor

(General) A rotor is the rotating part of an induction electrical motor. In x-ray tubes with rotation anodes the (usually copper) rotor is connected to a (molybdenum) stem rotating the anode (see the article Rotating anode). The rotating anode and the rotor are within the glass envelope (supported by ball bearings), and the stator producing the rotating magnetic field is fitted outside the glass envelope.

The mechanical speed of rotation of the rotor can be measured either in SI units in radians per second (rad/s) or in practical units of revolutions per minute (rpm). The speed of rotation of rotor in x-ray tubes can be usually varied between about 3000 rpm (normal speed) and 9000 rpm (rapid speed).

Related Articles: Rotating anode, Anode, Stator

RPA

(Radiation Protection) See Radiation protection advisor (RPA)

RPS

(Radiation Protection) See Radiation protection supervisor (RPS)
Safelight filter
(Diagnostic Radiology) See Darkroom

Sagittal plane
(General) To describe anatomical planes imagine a person standing in an upright position and dividing this person with imaginary vertical and horizontal planes. Anatomical planes can be used to describe a body part or an entire body.

Picture a vertical plane that runs through the body from front to back or back to front. This plane divides the body into right and left regions. This plane will pass approximately through the sagittal suture of the skull, and hence any plane parallel to it is termed a sagittal plane.

See Anatomical body planes

Sample volume
(Ultrasound) The sample volume is the volume from which pulsed wave spectral Doppler is investigated in the image in a duplex scanner. In conventional ultrasound scanners, the sample volume is usually displayed as a pair of parallel lines along the beam (Figure S.1). The position in the image can be altered with a trackball or trackpad. This changes its position on the screen and in turn allocates a specific depth and range from which Doppler sampling will be made. The indicated minimum length of the sample volume is usually around 1–1.5 mm, a practical limitation of the narrowband pulses used for Doppler. Larger sample volumes can be used; some systems offer sample volumes exceeding 15 mm. The sample volume is more complex than portrayed. No information is shown about its lateral extent which is dependent on the beam width. There is no information displayed about its dimensions in the elevation plane which is dependent on the slice thickness and which varies depending on the depth. Control of the sample volume is important for certain applications. A small sample volume allows the users greater precision as to where the sample volume is placed (Figure S.2). A large sample volume insonating the entire vessel is required if intensity-weighted mean velocity is to be measured in order to obtain Doppler signals from the range of velocities in the vessel.

The spectral display on Figure S.2 shows low velocities near the wall. In the left image, the sample volume is in the centre of the artery and low velocities are not detected or displayed.

Related Article: Pulsed wave Doppler

Sample volume effects in well-counter detectors
(Nuclear Medicine) This article refers to changes in well-counter geometric efficiency when counting different sample volumes. The geometric efficiency changes with source position within the well as seen in Figure S.3. It is highest at the bottom of the test tube and declines as the source approaches the well surface.

Consider two cases: (1) a sample with the radioactive substance diluted by water. The geometric efficiency in this situation decreases with increased sample volume. In case (2), a solution with constant activity concentration is poured into the test tube. In this case, the geometric efficiency initially increases linearly with sample volume but the proportionality is lost when the source volume approaches the well surface.

The sample volume significantly affects the geometric efficiency in the well counter, thus two samples should have the same sample volume in a comparative study. One approach, used when adequate sample volumes are available, is to fill the entire test tube because differences in total volume concentration will not affect the geometric efficiency to the same extent as differences in partially filled tubes.

Sample self-absorption and absorption in the test tube does not greatly affect the count rate unless the γ-ray energy is low (e.g. for 125I; 27–35 keV). To avoid any major sample volume effects, one should use identical test tube and sample volumes.

Related Article: Well-counter detector


Sampling theorem
(Magnetic Resonance) Sampling is the process by which a signal is converted from a continuous space of time or space into discrete space, for example an array of numbers. The sampling theorem states that if the maximal frequency in the continuous signal is f hertz (cps, cycles per second), then it can be completely determined from a discrete signal sampled with a spacing of (2f)−1. The sampling theorem is also known as the Nyquist–Shannon sampling theorem.

In MRI, the sampling theorem is important, for example when reconstructing an image from sampled k-space data. The read-out gradient and the size of the object determine the maximal frequency. The sampling distance in the k-space, Δk, then determines the FOV as FOV = 1/Δk. Foldover artefacts are due to violation of the sampling theorem.

Related Articles: Field of view, Nyquist criterion, Nyquist frequency, Nyquist theorem

FIGURE S.1 The axial length of the sample volume is displayed as two parallel lines in the centre of the artery (see arrow). The size of the sample volume is shown in the data fields (ringed) as 1.5 mm.
**SAR (scatter air ratio)**  
(Radiotherapy) See Scatter air ratio (SAR)

**SAR (specific absorption rate)**  
(Magnetic Resonance) See Specific absorption rate (SAR)

**Saturation**  
(Magnetic Resonance) For repetition times which are short relative to the $T_1$ of tissue there is little time for the net magnetisation vector to return to the $z$-direction between RF excitation pulses. For this reason, rapidly successive excitations lead to large reductions in longitudinal magnetisation and hence subsequently the MR signal. As TR reduces eventually excitation leads to almost no signal: this is saturation.

‘Saturation bands’ are often used in clinical MR to reduce signal (and thus reduce artefacts) from tissues adjacent to anatomy of interest. Saturation bands may be spatially selective or frequency selective.

Spatially selective saturation bands, for example may be used to avoid flow artefacts. In this case a 90° pulse is applied to tissue just outside the field of view and then the imaging sequence is applied immediately after. Blood flowing from the saturation band into the FOV has insufficient time to recover its longitudinal magnetisation, and so produces no MR signal.

Frequency-selective saturation bands operate slightly differently: relying on chemical shift. In order to apply a saturation band either to fat or water, the chemical shift between the two must be sufficiently large that an RF pulse can be used to selectively excite one and not the other and the main magnetic field, $B_0$, must be sufficiently homogeneous. Thus frequency-selective saturation bands are easier to implement at higher field strengths as the chemical shift is greater in absolute terms. The situation at 1.5 T is shown in Figure S.4.

Practically, a fat saturation band might be implemented as follows:

1. A 90° pre-saturation pulse tuned to the resonant frequency of fat is applied to the whole FOV, such that the bulk magnetisation vector of fat is flipped into the transverse plane.
2. A series of spoiler gradients are applied, to destroy the phase coherence of the fat signal.
3. The desired imaging sequence is applied immediately after the spoiler gradients, such that the fat protons have no time to recover their longitudinal magnetisation. The resulting images show signal from water nuclei only (Figures S.5 and S.6).

One limitation of the frequency-selective saturation band technique is that uniform saturation throughout the field of view requires excellent magnetic field homogeneity.

Furthermore, due to the extra RF pulse/gradients applied during each slice, sequences with either type of saturation band have increased SAR and increased acquisition time. Yet as saturation bands modify only the signal of the saturated tissue component (they do not alter other contrast characteristics) they can be applied to virtually any pulse sequence.

Saturation activity

(Nuclear Medicine) The creation of a radioactive source is a result of competition between the addition of new radioactivity by means of a given nuclear reaction and decay of the existing radioactivity. The competition will lead to an equilibrium state, establishing the saturation activity, when the rate of decay equals the rate of production.

The process is described by an exponentially increasing curve that will approach a constant value if the neutron flux or beam current is kept constant. (Figure S.7)

Consider the thin and thick activation formulas:

\[
A(t) = \frac{I \cdot \rho \cdot N_A \cdot \sigma \cdot x}{M} (1 - e^{-\lambda t}) \quad \text{and} \\
A(t) = \frac{I \cdot \rho \cdot N_A \cdot \sigma \cdot x}{M} \int_{E_{\text{threshold}}}^{E_{\text{end}}} \frac{\sigma(E)}{\partial E/\partial t} dE
\]

where

\[
I = \frac{dN_A}{dt}
\]

\(N_0\) is the total number of particles impinging on the irradiated area
\(\rho\) is the density
\(N_A\) is Avogadro’s number
\(\sigma\) is the cross section of each individual atom in a target with thickness \(x\)
\(M\) is the atomic mass of the target material. (Please see article entitled Activation formula thin target).

When the irradiation times goes to infinity or at least is large compared to the half-life of the radioactive product, then the factor \((1 - e^{\lambda t})\) approaches 1. This results in thin target and thick target saturated activities of

\[
A(t) = \frac{I \cdot \rho \cdot N_A \cdot \sigma \cdot x}{M} \int_{E_{\text{threshold}}}^{E_{\text{end}}} \frac{\sigma(E)}{\partial E/\partial t} dE \quad \text{and} \\
A(t) = \frac{I \cdot \rho \cdot N_A \cdot \sigma \cdot x}{M} \int_{E_{\text{threshold}}}^{E_{\text{end}}} \frac{\sigma(E)}{\partial E/\partial t} dE \quad \text{respectively.}
\]
Saturation curve

(Radiation Protection) Saturation curve is the resulting graph when the increase of one variable no longer produces an increase of the dependent variable. Usually the saturation curve has the shape of a hyperbola as shown in the Figure S.8.

An example is the current–voltage characteristics of an ionisation chamber.

Saturation voltage

(Radiation Protection) Saturation voltage is the voltage at which an observed effect does not increase further with increasing voltage. For example, in a gas-filled ion chamber and in the semiconductor diode detector the electric field should be high enough for assuring complete charge collection. The point at which there is no further increase in the collected charge is called the saturation voltage or saturation region.

Related Articles: Gas-filled radiation detectors, Germanium detectors, Ionisation chamber


Sawtooth voltage

(General) The saw-tooth voltage (or saw-tooth wave) is a voltage with non-sinusoidal waveform showing a slow linear rise and rapid linear fall (conventionally). It is named a sawtooth based on its similarity to the teeth on the blade of a saw. However, there are also sawtooth voltages in which the voltage ramps downward and then sharply rises. The latter type of sawtooth voltage is called a reverse (or inverse) sawtooth voltage.

The sawtooth wave is the form of the vertical and horizontal deflection signals used to generate a raster on CRT-based television or monitor screens.

Related Article: Cathode ray tube

Scan converter

(Ultrasound) When displaying images the monitor may use a different format than the data acquired during ultrasound scanning. A scan converter is a digital memory used to convert image scan data to that suitable for the monitor.

The line data will not normally fit in with the number of pixels in the memory. This is obvious for phased array and curvilinear array image formats where the spacing between the lines increase with depth (Figure S.9). The pixels between the line data will be filled in by new values using interpolation between the values in adjacent ‘line-data pixels’.

A number of other imaging processes are performed in the scan converter memory, for example averaging, freeze mode, read and write zoom and cine-loop.

Scan line

(Ultrasound) The scan line refers to the direction of the beam from which echoes are received and processed (Figure S.10). The term can be used for A-mode, B-mode and colour flow imaging; it tends not to be used for Doppler imaging where the term ultrasound beam is more commonly used.

Related Articles: B-mode, A-mode, Pulse echo

Scan projection radiograph

(Diagnostic Radiology) Scan projection radiograph (SPR) is the generic name used in CT scanning to describe the planning scan used prior to the CT acquisition. It is more commonly referred to by the trademark name used by particular CT manufactures, such as scanogram, scout view or topogram.

The SPR is acquired with the tube stationary, and the couch moving through the gantry. The usual position of the tube is either 0° or 180° (A-P/A-P SPR) (Figure S.11), or 90° or 270° (lateral SPR). The image obtained is similar in appearance to a conventional radiograph.
Scatter

Scatter refers to photons that undergo one or more Compton interactions prior to detection. Scattered photons add an uneven distribution of counts to the final image. These counts degenerate contrast and make the images unsuitable for activity quantification. The fraction of scattered events can in some cases (e.g. abdominal imaging) be as high as 60%–70%.

The probability for photon interactions (photoelectric, Compton and pair production) is proportional to the distance travelled in an absorbing medium. It also depends on the photon energy and absorbing medium characteristics (effective atomic number, density, etc.). When a photon undergoes a Compton interaction there is a change in direction.

Consider a source inside a patient at typical organ depth (~10 cm) while acquiring an image using a scintillation camera. In such a situation a photon could scatter either in the patient or in some of the detector parts (collimator). The event is assumed to have occurred somewhere along a perpendicular line to the detector face and the point of detection.

Because of the degenerative nature of scatter, several approaches to deal with it have been suggested, e.g. multiple energy window technique for SPECT and scatter estimation derived from a transmission image in PET.

Related Article: Automatic tube current modulation

Scatter air ratio (SAR)

Scattering is the re-radiation of acoustic waves with properties different from those of the incident wave. Opposed to the process of absorption, the energy is conserved in this process. Mathematically, scattering may be expressed a boundary value problem, where the scattered wave field is obtained using wave equations and matching boundary conditions at the surface of the ‘scatterer’. A scatterer, or scattering object, is one where the density and/or the compressibility of the medium are different from the surrounding medium.

To solve for the scattered pressure field, there are two principal methods. The first is to assume that the homogeneous wave equation is valid both inside and outside the scattering volume. The solution can be obtained by matching boundary conditions at the surface of the scatterer. In this case the geometry of the object must be known, and for instance, in this way the scattering from a sphere can be solved exactly.

For more complex problems, such as for scattering from tissues, a so-called Green’s function approach is employed. Here, a source term is added to the homogeneous wave equation. This leads to an integral equation in which the Green’s function is integrated over a volume including the inhomogeneities. The Green’s function can be seen as the wave arising from an elementary source. Basically, the terms of the integral equation then describes the effect of the source (transmitter) and the effect of the scattering objects.

Scatter air ratio (SAR)

(Spectroscopy) The scatter air ratio separates out the primary component of SAR from the total SAR to get the scatter contribution. It was defined by ICRU 24 as:

\[ \text{SAR}(d, A, E) = \text{TAR}(d, A, E) - \text{TAR}(d, 0, E) \]  

where \( \text{TAR}(d, 0, E) \) is known as the zero field tissue air ratio, which represents the attenuation of the primary beam in tissue with no scatter contribution. It can be measured using a column of water in narrow beam geometry, but more often is derived from extrapolation of the TAR values for finite field sizes (see Equation S.2).

The calculation of zero field TAR:

\[ \text{TAR}(d, 0, E) = \exp(-\mu_{\text{eff}} d) \]  

where \( \mu_{\text{eff}} \) is the effective attenuation coefficient of the photon beam energy \( E \).

There are three reasons that can explain this high fraction. The first in some cases (e.g. abdominal imaging) be as high as 60%–70%. There are three reasons that can explain this high fraction. The first reason is that only one of the two scattering photons might only deposit a small amount of energy. As an example; imagine an event where one of the annihilation photons is registered in a detector and the second photon is Compton scattered in the opposite detector crystal, thus depositing less than 511 keV, before escaping the crystal. Such an event would represent a true coincidence and a true LOR, but if the energy deposited does not exceed the lower limit of a potential energy discrimination window it will be ignored. As a result the pulse height analysis window must be widened to include these coincidences.

The second reason is that Compton scatter is the dominating interaction in scintillators at 511 keV and some incident photons might only deposit a small amount of energy. As an example; imagine an event where one of the annihilation photons is registered in a detector and the second photon is Compton scattered in the opposite detector crystal, thus depositing less than 511 keV, before escaping the crystal. Such an event would represent a true coincidence and a true LOR, but if the energy deposited does not exceed the lower limit of a potential energy discrimination window it will be ignored. As a result the pulse height analysis window must be widened to include these coincidences.

The third reason is the low-energy resolution in the LSO and BGO scintillators because of their low light output. The latter two reasons justify the use of a wide pulse height analyser window which at the same time provides an inadequate protection against scatter events. The energy discrimination window must be optimised in order to provide protection from scattered events but at the same time have a high sensitivity.

It is impossible to separate scattered events that originate from the body from the ones scattered inside the crystal, hence simple correction schemes such as dual-energy windows are not as successful for PET as for SPECT. There are two main approaches today for scatter correction.

In the first approach the correction is derived from the original scatter-contaminated image and a transmission image (representing the attenuation coefficient for different tissues). The transmission image is acquired using a transmission source or data from a CT scan. With data from these two images and some simple assumptions, a computer model can derive an estimation of the scatter distribution for each projection. The contribution of this scatter is then subtracted from each projection and the reconstruction is performed with scatter-corrected data. This method works well when the activation is contained inside the field of view (FOV). A source outside the FOV can lower the scatter correction accuracy. The method is computationally intensive.

The second approach involves examination of projection profiles just outside the object. After correcting for random coincidences it is assumed that all counts in such a profile are scatter induced. Scatter is assumed to be a low-frequency phenomenon with little structure and the scatter distribution can be extrapolated using a cosine or Gaussian function so that the scatter estimation covers the entire projection. The scatter distribution is then subtracted from each individual projection prior to image reconstruction. This method also accounts for activity outside the FOV. However, when scatter distribution is complex or when the object covers the entire FOV, this method can cause significant errors (e.g. 5% for brain imaging and more than 10% at the heart and lung interface).

Related Articles: PET, Compton scatter, Annihilation

A common approach to correct for scattered photons has been to collect projection profiles in a scatter window simultaneously with the photopeak window where the scatter window is placed below (in terms of energy) the photopeak window. The scatter projection profiles are multiplied with a weighting factor before subtracting them from the photopeak profiles. Which weighting factor to use depends on the size of the source, the energy window settings and the scintillation camera energy resolution, and this has to be determined experimentally. A limitation to this correction method is the fact that the scattered photons registered in the lower scatter window are more likely to have multiple scatter and as a consequence of this the spatial distribution of the events in the scatter window may differ from the scatter in the photopeak window.

A more useful method is based on two narrow energy windows placed adjacent to the photo peak window. The scatter in the photo peak energy window is estimated by the average of the acquired images in each of the narrow energy windows after correcting for differences in window width.

If the attenuation correction is performed prior to the scatter correction, then all events including scattered events will be corrected for. It is therefore important that scatter correction precedes the attenuation correction. Alternatively, one needs to reduce the magnitude of the attenuation coefficient to compensate for the extra counts in the image created by the scatter. The actual value of such an effective attenuation coefficient depends on the patient size, source location and tissue composition.

Iterative reconstruction methods sometimes may use the transmission images from the attenuation correction (see separate article) to analytically calculate the probability of scatter from a location in the patient being detected in a specific detector element.

**Related Article:** Attenuation correction SPECT


### Scatter factor

**(Radiotherapy)** The scatter factor SF for a field A is defined as the ratio

$$SF(A, E) = \frac{PSF(A, E)}{PSF(10, E)}$$

Hence it is the peak scatter factor normalised to 1 for a 10 × 10 cm² field, and quantifies the variation of scatter contribution with field size to the dose in tissue at the reference depth, normalised to the reference field. Note that $SF(10, E) = 1$ by definition.

This is also known as the phantom scatter factor, $S_p$, or the normalised peak scatter factor, NPSF.

**Related Article:** Peak scatter factor (PSF)

### Scatter phantom ratio (SPR)

**(Radiotherapy)** The scatter phantom ratio (SPR) bears the same relationship to tissue phantom ratio (TPR) as the scatter air ratio (SAR) does to the tissue air ratio (TAR). It separates out the primary component of TPR from the total TPR to get the scatter contribution.

The scatter phantom ratio (SPR) can be described as in Equation S.3. The calculation of scatter phantom ratio from tissue phantom ratios:

$$SPR(d, A, E) = TPR(d, A, E) - TPR(d, 0, E)$$

**Abbreviations:** TPR = Tissue phantom ratio, SAR = Scatter air ratio and TAR = Tissue air ratio.

**Related Articles:** Tissue phantom ratio (TPR), Scatter air ratio (SAR), Tissue air ratio (TAR).

### Scatter subtraction

**(Nuclear Medicine)** Scatter in an nuclear medicine image comes from the fact that the energy resolution of NaI (Tl) crystal generally is relatively poor and in order of 10% for 99m-Tc photons (140 keV). This means that in order to maintain reasonably large counting statistics, a large energy window (discriminator) is needed. Because a photon in a scattering event can lose only a small fraction of the energy and still change the direction significantly, there will be a chance that these photons will be registered within the energy window. These unwanted events will add to the total number of true events and will thereby reduce the image contrast because they appear to be coming from another location than the decay location. It is therefore desirable to correct for this scatter contribution. A scatter correction can be made by subtracting an estimate of the scatter in the main energy window by an estimate obtained either from an analytical procedure or from a measurement in a second energy window. The drawback with scatter subtraction is that if the main data is noisy and the scatter is noisy then the noise in the result from a subtraction will be even larger. Furthermore, in the case of inaccurate estimates and noise, a negative number can appear in the result that needs to be accounted for.

### Scattered radiation

**(Diagnostic Radiology)** Scattered radiation refers to all photons and charged particles resultant from scattering interaction between an incident photon or particle of ionising radiation and the medium.

Scattered radiation may travel in any direction away from the site of interaction. Therefore further ionisations and energy deposition in the medium may occur outside the primary radiation beam. Interactions between scattered radiation and the medium are important for radiation protection purposes – they are important in radiotherapy treatment planning when attempting to deliver the prescribed dose to the target volume (tumour) while minimising the peripheral dose to surrounding healthy tissues due to such scattered radiation.

Scattered radiation may also exit the medium altogether. Such transmitted scattered radiation is important in diagnostic radiology where it may be detected by the imaging receptor, contributing to the ‘fogging’ of the x-ray image. Scattered radiation is minimised in diagnostic radiology by the introduction of anti-scatter buxky grids between the patient and the image receptor.

**Related Articles:** Secondary radiation, Secondary electron, Peripheral dose, Bucky grids

### Scatterer

**(Ultrasound)** Scattering is the re-radiation of acoustic waves with properties different from those of the incident wave. As opposed to the process of absorption, the energy is conserved in this process. Mathematically, scattering may be expressed as a boundary value problem, where the scattered wave field is obtained using wave equations and matching boundary conditions at the surface of the ‘scatterer’. A scatterer, or scattering object, is one where the density and/or the compressibility of the medium are different from the surrounding medium. Scattering back to the transducer plays a large role in the formation of B-mode and Doppler images.

To solve for the scattered pressure field, there are two principal methods. The first is to assume that the homogeneous wave equation is valid both inside and outside the scattering volume.
The solution can be obtained by matching boundary conditions at the surface of the scatterer. In this case the geometry of the object must be known, and for instance, in this way the scattering from a sphere can be solved exactly.

For more complex problems, such as for scattering from tissues, a so-called Green’s function approach is employed. Here, a source term is added to the homogeneous wave equation. This leads to an integral equation in which the Green’s function is integrated over a volume including the inhomogeneities. The Green’s function can be seen as the wave arising from an elementary source. Basically, the terms of the integral equation then describes the effect of the source (transmitter) and the effect of the scattering objects.

**Related Article:** Green’s function

**Scattering, classic (coherent)**
*(Radiation Protection)* See Coherent scattering

**Scattering, coherent or Rayleigh**
*(Radiation Protection)* See Coherent scattering

**Scattering cross section**
*(Ultrasonography)* The scattering cross section is defined as the time-averaged total scattered power divided by the time-averaged incident intensity. The unit is in square meters, as implied by the name cross section. Physically this cross section corresponds to the area of the incident wave that contains the amount of power that is scattered by the scattering object. This means that the scattering cross section divided by the geometrical cross section of the object is a measure of how effectively the object scatters sound.

**Related Articles:** Absorption cross section, Extinction cross section, Differential scattering cross section

**Scattering, incoherent**
*(Radiation Protection)* See Incoherent scattering

**Scattering, Rayleigh**
*(Radiation Protection)* See Rayleigh scattering

**Scattering, Thomson**
*(Radiation Protection)* See Thomson scattering

**Scene-based registration**
*(Nuclear Medicine)* Scene-based registration, or projection-based registration, aims to register calculated projections from one 3D data set to calculated projection from another 3D data set. These methods can also apply to volume-to-image registration where projections of a 3D SPECT volume are registered to a planar scintillation camera image. By projecting the images onto the one-dimensional space the computational complexity is significantly reduced.

**Scilieren**
*(Ultrasonography)* Optical inhomogeneities in transparent material not visible to the human eye are named schlieren. These inhomogeneities can be visualised by the diffraction of visible light and can easily be seen in air above a lit candle or a hot surface.

Schlieren can be used to visualise the characteristics of a field from an operating ultrasonic transducer. The ultrasonic field will generate a variable pressure field which gives rise to variable refractive indexes in the field. Diffraction of light passing perpendicular through the field can then be used to visualise the ultrasonic field and to calculate the spatial pressure distribution (Figure S.13).

**FIGURE S.13** Schlieren image of the field from a single crystal ultrasound transducer.

**Scintillation camera**
*(Nuclear Medicine)* The scintillation camera is the most common imaging system used within the field of radionuclide imaging. It is also referred to as the Anger scintillation camera which is named after the inventor, Hal Anger. The basic principle of the scintillation camera is to depict the distribution of administered radioactive substances by ‘collecting’ emitted radiation.

The localisation and bio-distribution of the activity provides functional information about organ systems and individual organs, that is a high regional iodine uptake in the thyroid gland could sometimes be an indication of a pathological condition.

The four major components in a scintillation camera are a collimator, a NaI (Tl) crystal, typically, a light guide and an array of PM tubes. The collimator, which is a lead plate with holes, is a necessity in order to gain spatial resolution in the images. Photons with an oblique angle of incidence will be attenuated in the collimator whereas photons with a near parallel direction will pass through the detector and reach the crystal. Photons reaching the crystal will interact and through a number of processes create light photons. The number of light photons is proportional to the energy deposited in the interaction. The crystal must be a high attenuating material to register most of the incident photons. Choosing the right crystal with highest performance is an optimisation between a number of parameters, that is density, atomic number, light yield, decay time, sensitivity to mechanical stress, peak emission, etc. Light photons are transported through a light guide to the array of photomultiplier tubes (PM tubes). The use of an array of PM tubes, instead of single PM tube, makes it possible to localise the event using position logic circuitry.

The scintillation camera has a wide clinical use, for example whole body scans to localise cancer or metastases in the body and bone scans of the skeleton. Another use is for dynamic studies where a number of images are acquired during an examination. Gated acquisition is used to examine different phases of the cardiac cycle.

**Related Article:** SPECT


**Scintillation detector**
*(Radiation Protection)* The scintillation detector consists of a scintillator and a photomultiplier (PM) tube supplied with an external high voltage (Figure S.14). The scintillator, e.g. NaI (Tl), mounted in special container with reflective powder (MgO), is in optical contact.
with the PM tube. When an x-ray or gamma photon interacts with the crystal, its absorption results in the emission of optical radiation (with a maximum wavelength of about 420 nm) in all directions. The reflective powder protects the optical radiation photons from escaping the crystal. The light (the optical radiation photons) interacts with the photocathode by the photoelectric effect and its energy is transferred to the electrons. The quantum efficiency, QE, of this process is defined as

\[
\text{QE} = \frac{\text{number of photoelectrons emitted}}{\text{number of incident light photons}}
\]

and its value is 20%−30%. The PM tube multiplies the number of electrons thanks to the secondary electron emission from the system of electrodes (dynodes). The dynodes are held at the positive potential (the first of several hundreds volts) which accelerates electrons. The multiple stage multiplication for 10 stages (dynodes) is about 10^7.

The pulse height (amplitude) depends on the energy of the gamma radiation. The measured spectrum can be analyzed using a pulse height analyser (PHA), e.g. a single-channel analyser (SCA). Figure S.15 shows an example of a simple gamma radiation spectrum.

Solid state scintillation detectors are widely used for detection and quantification of gamma emitters in nuclear medicine, e.g. in gamma cameras, SPECT, and PET machines. Liquid scintillation counters are used to measure low-energy β^- and α-particles.

Scintillation detectors are very important in radiation protection, as their sensitivity is much higher than that of ionisation chambers and Geiger counters. They lack the ability to identify the type of radionuclide being detected, but they are ideal to quickly locate a lost gamma source or demarcate a contaminated area.

**Abbreviations:** MgO = Magnesium oxide, NaI (Tl) = Sodium iodide with thallium impurities, PET = Positron emission tomography and SPECT = Single photon emission computed tomography.

### Related Articles:
- Gamma camera, Liquid scintillation, Photomultiplier, Pulse height analyser, Scintillator

### Further Readings:

### Scintillator

*Nuclear Medicine* Radiation interacts with surrounding atoms and molecules by causing ionisation and excitation. As the atoms and molecules undergo recombination and de-excitation, energy is released, mostly as thermal energy. In a few materials, this energy is released as a number of low energetic photons (visible, UV). Material with this ability is called a scintillator. These scintillation materials are used in scintillation detectors.

Scintillation material can be divided into two groups: inorganic scintillators and organic scintillators.

Inorganic materials are solid crystals and the basic condition for scintillation comes from characteristics in their lattice structure. Individual atoms or molecules do not scintillate. In organic materials on the other hand the scintillation is a molecular property rather than an effect of crystal structure.

The general principle for all scintillators is that the amount of low energetic photons produced is proportional to the amount of energy deposited during the interaction.

**Inorganic Scintillators:** Many of the inorganic scintillators are impurity activated which means that an impurity, consisting of a small amount of atoms from a different element is added to the crystal structure. Sodium iodine (NaI (Tl)) (1948 Hofstadter) and caesium iodine (CsI (Tl)) are two examples of crystals with a thallium impurity. Most commonly used in SPECT is the NaI (Tl) scintillator. NaI (Tl) was first used for both SPECT and PET. Since NaI (Tl) has a long decay time and a low effective atomic number, it is unsuitable for PET imaging with its fast count rates and high-energy annihilation photons. Over time a number of new scintillators has emerged, i.e. BGO (1975 Nestor and Hung), GSO (Takagi and Fukazawa) and LSO (1990 Meicher). The individual properties of these scintillators are displayed in Table S.1. The most common scintillator used in PET today is LSO due to its high effective atomic number which allows it to effectively stop high-energy radiation.

**Photon yield** describes the number of photons produced per unit (keV) of deposited energy.

The emitted low-energy photons must pass through the crystal and into the photomultiplier tube where the low-energy photons will strike a photocathode. The number of photons passing or being reflected at this connection depends on the refraction index of the two materials. Similar refraction indices allow more photons to pass without being reflected back (where the reflected photon might be reabsorbed by the crystal).

Another important property, for both manufacturer and user, is that the scintillator is not sensitive to moisture or/and thermal and mechanical stress. NaI (Tl), for example is very fragile and also hygroscopic.

The electron yield at the photocathode is a function of the energy (wavelength) of the incident photons. For many photocathodes, the yield peaks at ∼400 nm. It is a desirable feature for a scintillator to have its peak emission in the vicinity of 400 nm.
**Organic Scintillators:** The scintillators in this group are in most cases liquid solutions, although there are some plastic organic scintillators. As previously mentioned the scintillation process of an organic scintillator is a property of individual molecules or atoms and not a lattice effect.

The radioactive sample is placed in a solvent containing scintillator material. Electrons freed in an ionizing interaction will transfer energy to scintillation molecules which will emit low-energy photons that are detected by one or more photomultiplier tubes.

**Related Articles:** PET, Photocathode, Photomultiplier tube, Photon yield, SPECT


**Scintillator**
(Radiation Protection) A scintillator is a substance in which molecules, absorbing the energy $E_a$ of photons (gamma or x-ray) or charged particles and neutrons, emit ultraviolet or visible light photons of energy $E_e < E_a$ (Figure S.16) with high efficiency. The emission is performed as a prompt fluorescence process with a decay time of a few nanoseconds.

There are two kinds of scintillators: inorganic and organic.

**Inorganic Scintillators:** A crystal of NaI (sodium iodide) activated with Tl (thalam), in quantities of about $10^{-3}$ mole fraction, to enhance the light output (photon emission) of visible radiation (415 nm), is an example of inorganic scintillators, in which the light efficiency is proportional to the radiation energy deposited in crystal. Figure S.17 shows a scheme of the role of the activator in an emission of a scintillation photon.

<table>
<thead>
<tr>
<th>Scintillator Material Properties</th>
<th>Density (g/cm³)</th>
<th>Effective Atomic Number</th>
<th>Decaying Time (ns)</th>
<th>Photon Yield</th>
<th>Index of Refraction</th>
<th>Hygroscopic</th>
<th>Peak Emission (nm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>NaI (Tl)</td>
<td>3.67</td>
<td>50</td>
<td>230</td>
<td>38</td>
<td>1.85</td>
<td>Yes</td>
<td>415</td>
</tr>
<tr>
<td>BGO</td>
<td>7.13</td>
<td>74</td>
<td>300</td>
<td>8</td>
<td>2.15</td>
<td>No</td>
<td>480</td>
</tr>
<tr>
<td>LSO (Ce)</td>
<td>7.40</td>
<td>66</td>
<td>40</td>
<td>20–30</td>
<td>1.82</td>
<td>No</td>
<td>420</td>
</tr>
<tr>
<td>GSO (Ce)</td>
<td>6.71</td>
<td>59</td>
<td>60</td>
<td>12–15</td>
<td>1.85</td>
<td>No</td>
<td>430</td>
</tr>
</tbody>
</table>

**TABLE S.1**
Properties of Different Scintillation Materials

![Figure S.17](image.png)

**FIGURE S.17** Scheme of a scintillation photon emission. $E_a$, energy gap.

Inorganic scintillators may be produced with different size and shape and are used as a gamma radiation detector in a gamma camera for medical imaging.

Examples of inorganic scintillators are: NaI (Tl), CsI (Tl), BGO, CdWO₄ for photon detection and lithium and boron glasses as neutron detectors. A novel scintillator is Lu(Y)AP, a crystal with better image resolution and reduced scan time and random noise. Its main application is expected in the next generation of PET machines, especially for pediatric care, cancer pathologies, heart and coronary artery diseases, neurological diseases.

**Organic Scintillators:** There are pure organic crystals (e.g. anthracen) and liquid scintillators composed of a scintillation solute (primary solute) such as e.g. PPO, a solvent, e.g. toluene or xylene, and often a secondary solute e.g. POPOP, called the wavelength shifter. Plastic scintillators may be shaped as fibres (Figure S.18), ribbons and thin films.

**Abbreviations:** BGO $\text{Bi}_4\text{Ge}_3\text{O}_{12}$ = Bismuth germinate, PET = Positron emission tomography, POPOP = Compound 1,4-bis-2-(5-phenyloxazolyl)-benzene and PPO = Compound 2,5-diphenyloxazole.

**Related Articles:** Gamma camera, Liquid scintillation (LS) counting, Scintillation detector


**Scoring**
(Nuclear Medicine) Semi-quantitative visual analysis was developed at Cedars-Sinai Medical Center for systematic and
reproducible interpretation of the myocardial perfusion images. Their approach uses three sets of slices from short axis views corresponding to the distal, mid-ventricular, and basal regions of the left ventricle as well as the apex visualised in a mid-ventricular long-axis image. A visual scoring system, based on either 17 or 20 segments is usually used.

The distal, mid-ventricular, and basal regions of the myocardium are divided into six segments each, while the apex is represented by two segments. Each of the segments is scored according to a 5-point scheme with the following definition:

- 0 is the normal
- 1 is the slight reduction of uptake or equivocal
- 2 is the moderate reduction of uptake, usually implies a significant abnormality
- 3 is the severe reduction of uptake
- 4 is the absence of tracer uptake

**Related Article:** Bullseye image


**Scout view**

(*Diasagnostic Radiology*) Scout view is a vendor name (GE) for the scan projection radiograph used in computed tomography.

**Related Article:** Scan projection radiograph

**SCR (silicon-controlled rectifier)**

(*Diasagnostic Radiology*) See Silicon-controlled rectifiers (SCRs)

**Screen film**

(*Diasagnostic Radiology*) A screen film radiographic receptor consists of a film in direct contact with either one or two intensifying screens. The function is illustrated in Figure S.19. The function of the intensifying screen is to absorb the x-ray photons and convert to light. Because of its composition and thickness the screen is an effective x-ray absorber. The film is more sensitive to light than to x-ray so the exposure to the film is ‘intensified’ by this process. Exposing the x-ray film both to the light from the screen and the x-rays from the beam increases the radiographic effect of the method, hence decreases the dose necessary for production of radiographic contrast.

**Screen selection**

(*Diasagnostic Radiology*) Intensifying screens are selected for specific clinical applications based on requirements for image detail and considerations for patient exposure. Two types of screens are illustrated on Figure S.20. Thin screens produce less blurring and provide better visibility of detail but do not absorb as much of the radiation as thicker screens. Thicker screens, described as being faster, can produce images with less exposure but with more blurring.

**Screen speed**

(*Diasagnostic Radiology*) Screen speed is a factor determining the sensitivity (or speed) of a radiographic receptor. It is the amount of exposure required to form image. Two screen speeds are illustrated on Figure S.21. A low-speed screen is generally thin to reduce blurring and absorbs only a fraction of the x-radiation. Therefore, a higher exposure to and through a patient must be used to form an image. Often the relative speed between various screen-film combinations can be expressed as 100, 200, 400, etc.

**Screen unsharpness**

(*Diasagnostic Radiology*) Blurring occurs within the thickness of intensifying screens as illustrated on Figure S.22. This is sometimes referred to as ‘screen unsharpness’ although unsharpness is just one of the effects of screen blurring. Two other effects are a reduction in visibility of anatomical detail and the other is a reduction in spatial resolution.

![Figure S.19](https://www.sprawls.org) - Screen film combination and effect. (Graphs courtesy of Sprawls Foundation, www.sprawls.org)

![Figure S.20](https://www.sprawls.org) - Effect on two types of screens on image resolution. (Graphs courtesy of Sprawls Foundation, www.sprawls.org)
Screen-film contact

(Diagnostic Radiology) See Film-screen contact

Sealed source

(Nuclear Medicine) A sealed, or closed source, refers to an encapsulated or otherwise contained radioactive source. From a radioprotection point of view, a sealed source is defined as a source that is unlikely to cause contamination.

Related Article: Unsealed source

Secondary barrier

(Radiotherapy) The design of a radiotherapy treatment depends on adequate shielding if the radiation exposure to those outside of the room is to be kept below the regulatory requirements. That part of the room that the primary beam falls on directly is called the primary barrier and will be the most heavily shielded part of the room (e.g. for a 10MV linear accelerator this will be of the order of 2.5 m of regular concrete).

Outside of this is a region where lower intensity radiation arising from scattered or leakage radiation falls need less in the way of shielding and the barrier here is known as the secondary barrier. The thickness of the wall in this region is of the order of 1–1.5 m depending on beam energy. See Figure S.23.

Related Article: Maze

Secondary circuit

(General) A secondary circuit is a circuit or winding (e.g. of a transformer) in which a current is produced by the electromagnetic induction of a current in a neighbouring circuit or winding called the primary circuit or winding. The primary circuit supplies the power that is ultimately used by the secondary circuit.

Related Articles: Transformer, High-voltage transformer, High-voltage circuit, Three-phase transformer

Secondary collimator

(Radiotherapy) Within the head of a linear accelerator once the electrons have impinged on the target and generated the x-rays to be formed into the treatment beam, a primary collimator is used to initially form a beam which may be used to treat a patient. Once the beam has been through the flattening filter a useable treatment field must be created. This typically takes the form of squares or rectangles which may be created by the secondary collimator. This secondary collimator is also commonly called the field-forming jaws, and the jaws are commonly termed x- and y-jaws. The four blocks consist of two to form the upper jaws, and two to form the lower jaws.

Typically the collimator is made from tungsten and is approximately 7 cm in thickness which attenuates the x-rays to the order of 1%–3%. Field sizes can be anything from a few millimetres up to a 40 x 40 cm² field as defined at the isocentre plane. It is important that the position of each jaw is known to within 1 mm at the isocentre plane. In older linacs the jaws only moved in pairs symmetrically about the central axis. However all modern linacs now allow each of the four collimator blocks to be moved independently. In this scenario at least one pair can then cross beyond the central axis by up to 10 cm in the isocentre plane. This allows so-called asymmetric treatments, where one of the beam edges is positioned coincident to the central axis. These are known as half-blocked treatments and are most commonly used in breast cancer treatments.

It is important that the secondary collimator provides a sharp beam penumbra irrespective of the field size. To achieve this linear accelerator manufacturers have created two options: (a) a straight edge is moved on an arc so that the collimator always presents the full surface to the diverging beam edge or (b) a curved edge is moved on a straight line (Figure S.24).

Related Articles: Collimation, Collimator, Dynamic wedge, Treatment head, Penumbra, Asymmetric jaws.
Secondary electron spectrum

(Radiation Protection) Secondary electrons are electrons given kinetic energy and liberated from atomic orbits as a result of interactions between incident photonic or charged particle radiation and the medium. These secondary electrons will have been given a range or spectrum of kinetic energies dependent on the energy and angle of the incident radiation, with some electrons having enough kinetic energy to cause further ionisations in the medium. The range of electron energies is referred to as the secondary electron spectrum, and will be characteristic of the type and energy of the incident radiation, and of the composition of the medium.

Related Articles: Secondary ionising radiation, Secondary electron

Secondary electrons

(Radiation Protection) Secondary electrons are electrons given kinetic energy and liberated from atomic orbits as a result of interactions between incident photonic or charged particle radiation and the medium. These secondary electrons may be given enough kinetic energy to cause further ionisations in the medium.

Related Articles: Secondary ionising radiation, Secondary electron spectrum

Secondary ionisation

(Radiation Protection) When secondary ionising radiation (photonic and/or charged particles) resultant from an interaction between an incident photon or particle of ionising radiation and the medium carries on to cause further ionisations in the medium, these further interactions are referred to as secondary ionisations. Secondary ionising radiation may travel in any direction away from the initial site of interaction. Therefore secondary ionisations and subsequent energy deposition in the medium may occur outside the primary radiation beam.

Related Articles: Secondary radiation, Secondary ionising radiation

Secondary ionising radiation

(Radiation Protection) Secondary ionising radiation refers to all photons and charged particles resultant from an interaction between an incident photon or particle of ionising radiation and the medium, which still have enough energy to cause further ionisations. Therefore secondary ionising radiation does not include any secondary photons or charged particles which are non-ionising, e.g. ultraviolet radiation.

Secondary ionising radiation may travel in any direction away from the site of interaction. Therefore further ionisations and energy deposition in the medium may occur outside the primary radiation beam. Such secondary interactions are important for radiation protection purposes – they are important in radiotherapy treatment planning when attempting to deliver the prescribed dose to the target volume (tumour) while minimising the peripheral dose to surrounding healthy tissues.

Related Articles: Secondary radiation, Secondary electron, Peripheral dose

Secondary radiation

(Radiation Protection) See Radiation, secondary ionising

Secondary standard

(Radiation Protection) A radiation detector (whether an ionisation chamber or solid state-based detector) used for day-to-day measurements of radiation output or dose is called a field instrument. The displayed reading on the instrument must have a traceable calibration such that it can be compared to any other instrument making the same measurement.

While the response of the field instrument may be checked regularly using a simple radiation check source of known output to ensure that the instrument is functioning correctly, it must be calibrated at less regular intervals (e.g. annually) against a primary standard held at national institute such as the National Physical Laboratory (NPL) in the UK, or the National Institute of Standards and Technology (NIST) in the USA. The primary standard used by such national institutes will be a radiation detector that is only used for calibration purposes, and is kept and used in a way to ensure that it has a very well-controlled and precisely defined response to given quantities of radiation.

It is not practicable for all field instruments used nationally to be sent to a single institute for direct comparison and calibration against the primary standard. Therefore, an intermediate step is introduced – the use of secondary standard calibration instruments. Such instruments are more widely available for calibration work, while not being used for day-to-day work as field instruments. A field instrument will be calibrated against the secondary standard which itself has already been calibrated against the primary standard, thus providing the necessary traceability to future readings on the field instrument.

Related Article: Reference ionisation chamber


Sector integration

(Radiotherapy) Sector integration is a method of determining dosimetric parameters for an irregular field from data for circular beams. It was first proposed by Clarkson (1941) and developed by Cunningham (1972) to calculate scatter air ratios for rectangular fields. In this method an irregular field is approximated by a number of circle segments as shown in the example in Figure S.25. The contribution from each sector is then integrated to give the total contribution at a point.


Secular equilibrium

(Nuclear Medicine) Transient equilibrium refers to the activity equilibrium that occurs in a decay series when the half-life of the parent nucleus is so long that the decay is negligible over the course of the observation period. The activity of the daughter nucleus is increased until it reaches an equilibrium state with the same activity.
as the parent nucleus. After a time period of 5 daughter half-lives the two activities can be said to be in equilibrium since $A_d = 0.98A_p$ ($A_d$ and $A_p$ is the activity of the daughter and parent nucleus, respectively) (Figure S.26).

**Related Articles:** Transient equilibrium, Bateman equation for secular equilibrium, Bateman equation for transient equilibrium

**Segmentation**

*(Nuclear Medicine)* Segmentation is an image processing tool used to reduce an image to its base components or objects. This is achieved by grouping all pixels that have certain defined characteristics. In nuclear medicine, for example image segmentation can be used to classify different types of tissue according to their tracer uptake. It is also useful in the automatic definition of regions of interest.

**(Radiotherapy)** Segmentation is a tissue structure contouring process in treatment planning. Normal organs of interest and target volumes are identified and outlined either automatically or manually on CT, MR and/or PET images for treatment planning and dose calculation.

**Selective excitation slice selection**

*(Magnetic Resonance)* See Slice selection

**Selenium detector**

*(Diagnostic Radiology)* The selenium detector or amorphous-selenium (a-Se) detector is an example of a direct conversion, flat panel detector used in medical imaging (x-ray) radiography. Selenium detectors can be used for both static radiography and real-time fluoroscopic imaging. It is referred to as a direct detection method as the image information is transferred from the incident x-rays directly to electrical charge with no intermediate stage via photoconductive detection. Figure S.27 shows a typical pixel of a selenium detector: It is formed from a continuous photoconductive

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**FIGURE S.25** Simple approximation of an irregular field using circle sectors.

**FIGURE S.26** The graph shows the build-up to a secular equilibrium. The half-life of the daughter is negligible compared to the half-life of the parent.

**FIGURE S.27** A direct selenium photoconductor detector pixel, the incoming x-ray creates an electron-hole pair and the liberated electron is then detected and stored by the detector element and storage capacitor.
layer, which is electronically coupled to an active matrix array. The active matrix array consists of a two-dimensional array of thin-film transistors (TFTs) which act like switches to read out the signal from each pixel. This detection method is in contrast to indirect detection where the incident x-rays are first converted into visible light wavelength photons by a phosphor screen prior to detection.

The top surface of the detector is constructed out of a continuous high-voltage electrode that allows the application of an electric field across the photoconductor, with the pixel elements held at a positive potential in relation to the top electrode. Detection occurs because x-rays that are absorbed by the photoconductor form electron-hole pairs, which are separated and migrated across the photoconductor to the opposite electrodes by the applied electric field. The electrons move towards the pixel electrode at the bottom of the detector and are stored as charge in the capacitor. The amount of charge stored is proportional to the energy of the x-rays incident upon the detector.

The photoconductive layer is usually constructed out of a thick layer of amorphous selenium (Z ≈ 34), as it easily deposits over large areas via traditional plasma deposition, has a low dark current, and good electron and hole transportation properties. The thickness of the selenium used in the photoconductor is dependent on the x-ray absorption depth (δ), the imaging application and the K and L edge energies. For mammography with mean peak energy 20 keV, the a-Se thickness required is δ, about 100μm while for chest and general radiography with 60 keV it is 2000μm.

To detect an image, many pixels form an active matrix array shown in Figure S.28. In this each pixel is formed by a charge collection electrode, a storage capacitor and a thin film transistor switch/gate. To detect the image signal, the charge stored in each pixel’s capacitor is drained by the charge collector electrode and stored as charge in the capacitor. The amount of charge stored is proportional to the energy of the x-rays incident upon the detector.

The resolution of direct conversion detectors are almost entirely dependent on the pixel aperture and have a higher resolution than indirect-conversion detection for the equivalent pixel size as the electric field forces the electrons to travel directly to the electrode without lateral diffusion. Another important advantage of direct over indirect detection is the pixel fill factor, that is the percentage of the pixel that is sensitive to incident x-rays, defined as

\[
\text{Fill factor} = \frac{\text{Light sensitive area}}{\text{Full area of pixel}}
\]  

(S.4)

By appropriate design of the electrode the electric field can be adjusted to increase the effective fill factor of each pixel as electrons that are liberated above insensitive areas of the pixel can be redirected to active region. This allows fill factors of 85%–95% to be common.

**Abbreviation:** TFT = Thin film transistor.

**Related Articles:** Amorphous selenium, active matrix array, thin film technology, electron-hole pair

**Further Readings:** Amorphous selenium, active matrix array, thin film technology, electron-hole pair

**Self-absorption**

(Radiation Protection) Self-absorption is the absorption of radiation (emitted by radioactive atoms) by the material in which the atoms are located; in particular, the absorption of radiation within a sample being assayed.

The scattered radiation produced when a radiation beam impinges on a medium is partially absorbed within the medium. This phenomenon may be described as self-absorption as well.

A third example of self-absorption occurs in scintillation detectors. The incident ionising radiation is converted to light photons within the scintillator; however a fraction of these light photons are absorbed within the scintillator itself and are not detected by the photomultiplier optically coupled to it.

**Related Articles:** Radiation absorption, Scattered radiation

**Self-shielded cyclotron**

(Nuclear Medicine) A self-shielded cyclotron is a low-energy cyclotron in which the steel frame or yoke serves as primary radiation shield. Additionally hydraulically driven movable blocks made of specially formulated concrete surround the cyclotron and add more radiation shielding.

**Semiconductor detector**

(Radiation Protection) A semiconductor detector of ionizing radiation is a solid state detector in which electron-hole pairs are created when charged particles (e.g. alpha) or photons (gamma, x-rays) pass through the detector. Figure S.29 shows the electron energy band structure for: (a) intrinsic, (b) p-type and (c) n-type semiconductors.

The density of solid state semiconductors is more than 1000 times greater than that of a gas, and the energy needed for the creation of one pair electron-hole (an electric charge carrier) is not greater than 3eV (very small in comparison with the 100eV which is needed for scintillation detectors or with the approximately 35 eV
Semiconductors

**Related Articles:** Absorbed Dose, Air Kerma, Exposure, Cylindrical ionisation chamber

**Further Reading:** PTW, 2008/2009. Ionizing Radiation Detectors, p. 22.

**Sensitivity (Nuclear Medicine)** The sensitivity is a measure of how many of the emitted photons are being detected by the PET system. This sensitivity is determined by a number of factors, primarily the attenuation efficiency of the scintillation material and the geometric parameters. For a source located inside an absorbing medium, the true coincidence rate \( R_{\text{true}} \), for two coincidence detectors is given by

\[
R_{\text{true}} = E\varepsilon_0 \gamma_{\text{ADC}} e^{-\mu T} \tag{S.5}
\]

where
- \( E \) is the source emission rate
- \( \varepsilon_0 \) is the intrinsic efficiency of the detectors (the fraction of incident photons detected)
- \( \mu \) and \( T \) are the linear attenuation coefficient and the thickness of the patient, respectively
- \( \gamma_{\text{ADC}} \) is the geometrical efficiency of the detector, that is the number of annihilation events where both photons are emitted in a direction where they can be detected by two opposite detectors

The geometrical efficiency is maximised if the source is located at the middle of a centreline between two detectors. When the source is moved away from the centre point the efficiency decreases. A more appropriate measure for distributed sources is the *average geometric efficiency*, given by

\[
\frac{\gamma_{\text{ADC}}}{D} = \frac{2A_{\text{avg}}}{3\pi D^2} \tag{S.6}
\]

where
- \( D \) is the distance between the detectors
- \( A_{\text{avg}} \) is the area of the individual detector

**Related Articles:** PET, Beta decay


**Sensitivity profile (Nuclear Medicine)** In nuclear medicine, sensitivity profiles are used to evaluate different parameters in imaging equipment. In PET imaging the sensitivity, i.e. events per unit time, depends on the PET scanner, the data acquisition type (2D or 3D acquisition) and the source position along the axial direction. The sensitivity

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**Semiconductor**

**(General)** A semiconductor is a substance that can conduct electricity under some conditions but not others. These properties of such substances turn them into a good medium for the control of electrical current. Conductance varies depending on the current or voltage applied to a control electrode, or on the intensity of radiation.

The specific properties of a semiconductor depend on the impurities, or dopants, added to it. Two types of dopants can be used having either an extra peripheral electron \((z + 1)\) for N-type or a lack of one electron \((z - 1)\) for P-type regarding the number of peripheral electron \((z)\) of the semiconductor used. So, an N-type semiconductor carries current mainly in the form of negatively charged electrons. A P-type semiconductor carries current predominantly as positive electric charge–holes.

Some of the most widely used materials for manufacturing of semiconductors are antimony, arsenic, carbon, germanium, selenium, silicon, gallium arsenide.

**Semiflex chamber**

**(Radiation Protection)** The semiflex chamber is a cylindrical ionisation chamber used for in vivo dose measurements in external beam radiotherapy and brachytherapy. The chamber can be used for absorbed dose to water, air kerma and exposure measurements.
can be measured by acquiring images of a line source oriented along the central axis. Another example of a sensitivity profile is the spectral sensitivity of the photocathode in a photomultiplier tube. The photocathode spectral sensitivity must have an overlap with the corresponding scintillation light emission spectra in order to give a signal enhancement.

**Sensitometer**

*(Diagnostic Radiology)* A sensitometer is a light-emitting instrument for exposing areas on a film to a range of exposure values, usually in a step pattern as illustrated in Figure S.30. The colour of the light has to match the sensitivity spectrum of the film.

Films exposed with a sensitometer can be used to determine the characteristics of both the film and the film processing.

**Sensitometry**

*(Diagnostic Radiology)* See Sensitometer

**Septal penetration**

*(Nuclear Medicine)* Photons emitted with an oblique angle relative to the collimator holes can penetrate the septal wall before being detected. This problem is referred to as *septal penetration* and it gives rise to degradation in spatial resolution.

The probability of septal penetration is proportional to photon energy and inversely related to atomic number of the collimator material and septa wall thickness. For a parallel-hole collimator a registered photon is assumed to have been emitted somewhere along an event line perpendicular to the detector surface. If the photon has penetrated the septa the event will be misplaced. The misplacement is relatively short since the probability of a photon penetrating multiple septa walls is relatively small.

Septal penetration leads to a degradation in spatial resolution but the effect is only prominent when using high-energy γ emitters and high-resolution collimators with thin septa walls.

**Related Articles:** Collimators, Parallel-hole collimator


**Septum**

*(Nuclear Medicine)* A barrier consisting of a high attenuating material used to shield detectors or attenuate radiation with an oblique angle of incidence.

One example of septa are the walls separating the holes in a collimator. The thickness of the wall is designed for a certain purpose, i.e. thick walls for high-energy photons and thinner walls for low-energy photons. Another use for septa walls is when separating detector rings in a PET scanner.

**Sequestration targeting**

*(Nuclear Medicine)* Sequestration refers to the process where the spleen identifies and removes damaged red blood cells. This is used when studying the spleen. Small portions of the patients’ blood are removed and labelled with radioisotopes. Before the cells are reintroduced into blood circulation they are heated to cause cell damage so that the spleen will gather the radiolabeled cells.

**Serial exposures**

*(Diagnostic Radiology)* Serial exposures (also known as image sequences) are images obtained in a rapid succession (series). These are used for recording dynamic processes in rapidly moving organs (as heart). The series can be with speed from 2–8 images/s (when using record on x-ray film) to 25–30 or more fps (in digital x-ray systems). See the articles on Image sequences and Pulsed fluoroscopy.

**Related Articles:** Image sequences, Pulsed fluoroscopy

**Serial organs**

*(Radiotherapy)* Radiation treatment inevitably affects normal tissue and so may cause radiation-induced adverse effects. The tolerance of normal tissues to radiation depends on the ability of the clonogenic cells to maintain a sufficient number of mature cells suitably structured to conserve organ function. The tissue architecture is thought to be important in determining the tolerance dose for partial organ irradiation.

In radiotherapy, it is generally the case that the total dose that can be tolerated depends on the volume of tissue irradiated – the dose-volume effect. It has been suggested that groups of cells within an organ are organised into collective bodies called functional subunits (FSU). The arrangement of the FSUs within the tissue is thought to be an important factor in determining the volume dependence of an organ. In serial organs, the FSUs are arranged in series, like the links of a chain, and the integrity of each is critical to organ function. Damage to a single FSU is sufficient to cause a complication. Radiation damage to such tissues is expected to show a binary response: normal function for doses below a threshold dose above which there is loss of function. For these tissues, the greater the volume of tissue irradiated, the steeper the sigmoid dose–response curve becomes and the threshold dose decreases. The effect of increasing the irradiated volume is greatest with changes to small volumes: Once a large number of FSU’s are irradiated any further increase in volume has little effect on the threshold or slope of the sigmoid dose–response curve. The steepness of this curve for serial organs means that they are sensitive to small increases in ‘biologically effective’ dose. Therefore administering large dose fractions to the spinal cord, for example will increase the biological effectiveness per unit of dose so that using a large volume together with large dose fractions, as occurs in palliative treatments, could augment the danger of myelitis.

When planning radiotherapy treatment, assessment is always made of the dose to normal tissues and modern commercial treatment planning systems have a number of tools to aid this process including dose-volume histograms (DVH). For serial organs, serious complications are likely to be dominated by small-volume, high-dose effects so particular attention is paid to the maximum dose they receive.

Examples of serial organs are spinal cord, optic chiasm, optic lens, small bowel. It should be noted however that most real normal tissues have a mixed parallel and serial architecture.
Abbreviations: FSU = Functional sub-unit and DVH = Dose-volume histogram.

Related Articles: Adverse events, Biological effective dose, Dose response model, Dose volume histogram, Parallel organs, Sigmoid dose–response curve, Tolerance


Servomotor

(General) See Servomotor

Servomotor

(General) Servomotor is the motive element in a servomechanism. A servomechanism (or servo) is an automatic device that uses error-sensing feedback to correct the performance of a mechanism. A common type of servomechanism provides position control using an electric servomotor. Usually, servomechanisms operate on the principle of negative feedback, where the control input is compared to the actual position of the mechanical system as measured by some kind of transducer at the output. Any difference between the actual and expected values is amplified and used to drive the system in the direction necessary to reduce or eliminate the error.

Related Article: Voltage regulation

Set-up error

(Radiotherapy) The set-up error accounts for the variability in positioning the patient with respect to the treatment beams. Errors are either systematic, geometric errors that arise during treatment preparation and are effectively ‘frozen’ into the patient’s treatment, or random treatment execution errors. 

Causes of systematic error include

- Linac geometry error (laser error, collimator axis error, or beam alignment error)
- Treatment planning system error (volume growing, image transfer)
- Geometric imaging error (CT couch indication, CT laser error)
- Delineation uncertainty
- Systematic set up error (patient position)
- The random variability in patient positioning inherent in the scanning process used to prepare the treatment

All the above errors should be minimised as much as possible with regular QA and training. Additionally, the extent of systematic errors can be limited by setting an action level for correcting discrepancies revealed by portal imaging. The remaining error must be accounted for by the use of planning margins.

Margins can be quantified by multiplying the variance of the error distribution by a multiplication factor according to the level of confidence required to encompass the entire CTV. The variance of the error distribution should include contributions from all the above sources, combined in quadrature if Gaussian. Breathing positional errors are not Gaussian in nature.

Various numerical simulations have been carried out to determine the multiplication factors and treatment margins necessary to achieve a certain probability of target coverage as a function of the size of the two types of error. These reveal the effects of systematic errors to be of order 3 times more important than random errors.

EXAMPLE: MARGIN FORMULAE

Stroom et al. derived the follow margin formula for 99% of the CTV to receive 95% dose:

\[ M = 2\sigma + 0.7\sigma \]

Van Herk et al. calculated margins for a minimum CTV dose of 95% in 90% of patients:

\[ M = 2.5\Sigma + 0.7\Sigma \]

where

- \( M \) is the margin size
- \( \Sigma \) is the size of the systematic error and
- \( \sigma \) is the random error

It is also important to remember that set-up error will also effect the position of organs at risk.

Abbreviations: IM = Internal margin, SM = Set up margin and EPID = Electronic portal imaging device.

Related Articles: IM internal margin, SM set up margin, Electronic portal imaging device EPID


Set-up margin

(Radiotherapy) A margin needs to be added to the clinical target volume (CTV) to form the planning target volume (PTV) to account for any positional errors from the planning information. The ICRU Report 62 (ICRU 1999) divided this margin into the set up margin (SM) and internal margin (IM), in order to separate out the contributory sources of positional error into set up error and physiological error, respectively. The SM accounts for variations in patient positioning and alignment of the treatment beams. Immobilisation devices and portal imaging help to reduce the SM by improving the reproducibility of patient position. Regular QA helps to reduce the SM by improving the precision and accuracy of the treatment beam motion.

The remaining source of error, i.e. the physiological variation of the tumour size and shape is accounted for by the internal margin. The CTV plus the IM defines the internal target volume (ITV). The total required margin (SM + IM) can be quantified from the standard deviations (\( \sigma \)) of the associated probability distributions. If the error distributions can be treated as Gaussian and independent, then their standard deviations should be combined in quadrature (i.e. sum of variances), with a multiplication factor according to the level of confidence, as shown in Equation S.7.

In other words, the total margin would be proportional to the combined standard deviation.

Combining margins for set up error and physiological (internal) error:

\[ \text{Total margin} \propto \sqrt{\frac{\sigma_{SM}^2}{2} + \sigma_{IM}^2} \]  \hspace{1cm} (S.7)
Severity

(Nuclear Medicine) The severity of the abnormality indicates how low the tracer uptake with the abnormality is and usually is expressed as the sum of SD (standard deviation) below the lower limit of normal. Extent and severity are two important parameters for assessment and management of coronary artery disease (CAD).

Related Article: Extent

Shading artefact

(Magnetic Resonance) The shading artefact appears as a variation in signal intensity on the image. The main cause of this artefact is the uneven excitation of nuclei due to an RF pulse with a flip angle other than 90° or 180°. Other causes of shading are the uneven loading of the coil, which could occur if the patient is touching the coil or inhomogeneities within the magnetic field, which can be improved with shimming.

Abbreviation: RF = Radiofrequency.

Shadowing

(Ultrasound) Shadowing is an artefact effect where the region behind a structure with high attenuation is interrogated with a beam with decreased intensity, with the consequence that, in B-mode (imaging) ultrasound the tissue deep to this appears with reduced brightness compared to adjacent similar tissue (Figure S.31). In colour Doppler flow imaging the consequence of shadding is that colour signal is lost for regions deep to highly attenuating or highly reflecting tissue such as calcified plaques. The combination of bright echoes with corresponding shadowing is useful for detection of gallstones and arterial plaques, for instance.

Shadowing might also occur due to scattering by gas bubbles which, in turn, might have been created by cavitation due to the high-intensity interrogating ultrasound itself (Leighton, in Duck et al. 1998).

In transverse section on Figure S.31 there is strong reflection from the graft surface (grey arrow). There is little or no transmission onwards through to deep tissue and there is a corresponding dark shadow in the image (white arrow).

Related Articles: Attenuation, Acoustic Impedance


Sharpness

(Diagnostic Radiology) Sharpness is a visual characteristic of an image where objects and structures appear to not be blurred. Unsharpness is when there is a perceived effect of blurring. Better image sharpness assumes better spatial resolution.

Shear waves

(Ultrasound) Shear waves, or transverse waves, are characterised by particle motion perpendicular to the direction of propagation. These can propagate easily through some solid materials, such as steel and bone, but they do not effectively travel in soft tissue. Some applications however have been reported where transverse waves are generated, and then detected, but normally, only longitudinal waves are considered in ultrasound imaging.

Shell model of nucleus

(General)

Nuclear Structure: The atomic nucleus consists of neutrons and protons commonly referred to as nucleons. The nucleons belong to the family of baryons, composite particles made of three quarks, and the baryons belong to the hadrons, i.e. all particles made of quarks. The hadrons are bound states of quarks, held together by the strong force. The nucleons in the atomic nucleus are bound together by the nuclear force (the residual strong force), which is the residue strong interaction between the quarks that constitute the nucleons.

In order to describe nuclear structure, a number of models have been developed. The liquid drop model, where the nucleus is regarded as a drop of neutrons and protons, was one of the first models that successfully described the basic principles of the nuclear binding energy (semi-empirical mass formula). However, there were systematic deviations not accounted for in this model. These deviations could be explained if the nucleons in the atomic nucleus exhibited a shell structure ‘similar’ to the situation for the atomic electron.

Nuclear Shell Model: The atomic nucleus is a quantum n-body system, where the nucleons are bound together by the residual strong force, the nuclear force. The nucleons are fermions and they obey the Pauli exclusion principle stating that two fermions cannot occupy the same quantum state. The nucleons are according to this principle not interacting even if they are closely packed, as there are no states available for the collision processes. Experimental evidence, like magic atomic numbers, suggests a nuclear shell structure, as mentioned above. In other words, each nucleon seems to move inside a potential well created by the forces from the all
the other nucleons in the nucleus. The nuclear n-body problem is thus replaced by n single-body problems. This independent particle model, based on the concept of a mean field, has been very successful in explaining nuclear structure.

In order to solve the single-body problem, the potential of the mean field must be defined. A historically successful phenomenological method was proposed by S. G. Nilsson (Nilsson 1969), where a deformed harmonic oscillator together with an empirical spin-orbit coupling was used to characterise the nuclear potential. Today, with powerful computers available, self-consistent methods (Hartree-Fock methods) are used to deduce the nuclear potential from the nucleon-nucleon interactions.

The nuclear shell model, correctly describing magic numbers, has been able to predict nuclear properties based on the single particle energy level scheme.


Shielded gradients
(Magnetic Resonance) Switching of gradients induces voltages and currents in conducting material throughout the system, called eddy currents. These eddy currents produce fields that oppose the applied gradients. In order to minimise eddy currents, gradient coils can be shielded by means of applying a secondary coil pair at a larger radius than the main gradient coil. Current flows in the opposite direction to the gradient coils which produces fields that counteract the primary fields. The shielding leads to a reduction of the primary field, thus a larger current is needed to obtain the desired gradient field.

Related Articles: Eddy currents, Shielding


Shim coils
(Magnetic Resonance) The purpose of shimming is to compensate minor spatial inhomogeneities in the $B_0$ magnetic field in the imaging object. This is done by imposing $B_0$ fields of specific spatial dependence, each of which is created by a shim coil. These are either superconducting (static) or resistive (mounted within the magnet bore). Common in MRI systems is a set of 9 coils creating additional $B_0$ fields of linear or quadratic spatial dependence. These fields conform to spherical harmonic functions, which are adjusted automatically before every examination, because the shape and dielectric properties of the body/sample imposes distortions of the $B_0$ field.

Related Article: Shim coils


Shimming
(Magnetic Resonance) Even at the isocentre of the magnet, the static field ($B_0$) is not perfectly homogeneous. The term ‘shimming’ denotes the process of reducing the spatial inhomogeneities over a certain region of interest. These are given in ppm units and form part of the system specification.

Passive shimming is performed during installation by distributing pieces of iron in the bore of next to the cryostat. Active shimming involves adjusting DC currents in a set of (up to 18) shim coils creating additional $B_0$ fields of linear or quadratic spatial dependence. These fields conform to spherical harmonic functions, which are adjusted automatically before every examination, because the shape and dielectric properties of the body/sample imposes distortions of the $B_0$ field.

Related Article: Shim coils


Shock absorber
(General) The function of a shock absorber or ‘damper’ is to reduce or minimise sudden mechanical movement or vibration of a device which may otherwise cause the device to malfunction or sustain physical damage.

Shock absorbers are usually mechanical in form, based on absorbing the energy of an applied impulse by gradual dissipation through friction between surfaces or by viscous flow of some enclosed fluid such as oil or air within a piston.

Shock waves
(Ultrasound) A shock wave is an abrupt increase in acoustic pressure, which is built up as a result of non-linear propagation. The limiting case of a N-shaped waveform is termed a shock wave, and here the harmonic components in the spectrum have amplitudes that fall off by $1/n$, where $n$ is the nth harmonic. The higher harmonics are quickly attenuated, however, and the N-shaped waveform cannot be sustained. After some additional distance, the waveform will resemble a sinusoid but with much smaller amplitude than it originally had.

The underlying mechanisms are, first, a convection effect, and second, an effect of nonlinear compressibility. Both of these effects have the result that pressure peaks travel faster than rarefactions, and will thereby eventually encroach upon them. The effect is more pronounced the higher the acoustic pressure.

The generation of harmonics generated from a shocked waveform are used in harmonic imaging, for instance. Shocked waveforms are also encountered in lithotriptors for the disintegration of kidney stones. Here, the peak acoustic pressures can reach 100 MPa.

Short circuit
(General) A short circuit is a condition in the electrical system where energised conductors come in contact (or generate an arc by coming in close proximity) with each other or with ground, allowing (typically large) fault currents to flow.

Short tau inversion recovery (STIR)
(Magnetic Resonance) Short inversion time (TI) or short tau (τ) inversion recovery, STIR, is a signal suppression or nulling technique with an specific inversion time, giving zero signal for components with longitudinal relaxation time TI. Usually this technique is applied to suppress the fat signal, whereas for nulling of water/CSF (fluid attenuated inversion recovery [FLAIR]) a long TI is required.

The concept of nulling can be understood by inspection of the signal curves obtained as function of the selected TI value in an IR sequence (see Figure S.32 below and Inversion recovery). At a specific TI value, which depends upon TI of the tissue as well as the selected TR, the signal will be zero for phase representation as well as for magnitude (or modulus) representation of the IR image.

Short tau inversion recovery (STIR)

For very long TR values, $T_{null} = 0.69T_1$ indicating $T_{null}$ values between 150 and 200 ms for STIR and above 2000 ms for FLAIR.

In the diagrams to the left in Figure S.32, signal (arbitrary units) is plotted against inversion time (ms) for three different tissue types having low T1 (green lines), medium T1 (red lines) and high T1 (blue lines). In the images to the right, brain image from a healthy volunteer at 1.5 T are shown for three selected TI values.

Note specifically the modulus representation at TI = 150 ms, where TI is close to the value optimal for nulling of fat (STIR) and at TI = 2200 ms which illustrates a FLAIR type sequence. Note also that, neglecting transverse relaxation effects, optimisation of TI for STIR and FLAIR depends upon field strength (via T1 dependence) and the selected TR. Furthermore the inversion times given above assume that the longitudinal magnetisation fully recovers before applying the next inversion rf-pulse.

The STIR sequence is often combined with a fast spin echo (FSE) imaging sequence (see Figure S.33). This combination is called fast inversion recovery or turbo inversion recovery. Fast IR

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**FIGURE 5.32** (See colour insert.) Illustration of the concept of nulling. (Figure courtesy of Sara Brockstedt, University of Lund, Sweden.)

**FIGURE 5.33** (See colour insert.) Conceptual illustration of a fast inversion recovery pulse sequence.
sequences have opened the possibility to perform the inversion recovery experiment (see *Inversion recovery*) within reasonable acquisition times, since in this case $T_{\text{aq}}$ will be reduced by the echo train length (ETL) factor as in FSE.

Additional contrast can be obtained in STIR and FLAIR by combining a high TR value and an adequately chosen TI value with a relatively long TE. As an example, in the modulus representation of STIR, the low inversion time chosen for nulling of fat enforces mirroring of signal values for most tissue types in the brain. Hence, tissue with high-T1 values will have higher signal than tissue with lower T1. In cases where the high-T1 tissue also has higher T2 than the low-T1 tissue, the choice of a long echo time will further enhance image contrast (Figure S.34). In Figure S.34a, the vertical dashed line shows the selected TI value for nulling of fat. Using this TI, a tissue type with high T1 (blue) will give higher potential signal intensity than a tissue type with medium T1 (red). If the transverse relaxation time $T2$ is higher for the ‘blue’ tissue than for the ‘red’ tissue, the $T2$ decay during TE will give a measured signal that further enhances signal intensity differences between the two tissue types. A selection of TE/TR/TI as described can be used to enhance signal differences between liquid and brain matter, but also to enhance the more subtle signal differences between grey and white matter.

To the right, a fast IR sequence with parameters $TE = 60\, \text{ms}$, $TR = 4900\, \text{ms}$ and $TI = 150\, \text{ms}$ (STIR type with additional T2 contrast) is shown (Figure S.34b).

**Related Articles:** Echo train length, Fluid attenuation inversion recovery (FLAIR), Inversion recovery, Inversion time, RF pulse

**Shortest exposure time**
(Radiotherapy) See Exposure time

**Shrapnel**
(Magnetic resonance) The term shrapnel is often used to describe fragments or shot intentionally included in an explosive device but in a MR by shrapnel it is indicated a metallic fragment in the patient’s body. The internal presence of metallic fragments in patients is an exclusion criterion for the MR safety before the MRI examination, even if the fragment is small in dimension. Patients who have been injured by bullets or shrapnel or who work with metals could have internally metallic fragments. The main concern is fragment located in or around eyes as the static magnetic field exerts a force on ferromagnetic objects and the metal fragment in the eye could move or be displaced and could result in injury to the eye or the surrounding tissues. If the patient works or has worked with sheet metal or as a welder it is probable that he has metallic fragment or slivers placed in or around the eye and his line of work should be specifically questioned to determine if a MRI examination is possible. The presence of small foreign bodies could be assessed by a plain film radiography that is considered an acceptable standard in screening for intra-ocular metallic foreign bodies with sufficient size to cause ocular damage. Furthermore, a CT scan of the orbits could determine more accurately the presence of smaller fragments.

**Shunt**
(General) In electronics, a shunt is a device having appreciable impedance connected in parallel across another device and allowing some part of the current to pass around to another point in the circuit. An ammeter shunt allows the measurement of current values too large to be directly measured by a particular ammeter. The term shunt is used in filter circuits to refer to the components connected between the line and common. Capacitors can be used as shunts to redirect high-frequency noise to ground. Where devices are particularly sensitive to reverse polarity of signal or power supply, a Zener diode may be used as a shunt to protect the circuit.

**Related Articles:** Zener diode, Voltage limiter

**Shutter**
(Diagnostic Radiology) See Cineradiography

**SID (source-to-image distance)**
(Radiotherapy) See Source-to-image distance (SID)

**Side lobes**
(Ultrasound) Due to interference, sound is emitted in a number of principal directions from a disc source. Perpendicular to the disc, along the central axis, is the main lobe, which contains the most energy. However, in the direction that corresponds to the divergence

![Image](Image 371x117 to 514x295)

**FIGURE S.34** (See colour insert.) Illustration of a modulus representation of a STIR sequence, where additional contrast is allowed by using a long TE: (a) Variation of signal with echo time for STIR pulse sequence and (b) fast inversion recovery of the brain (TR = 4900 ms, TI = 150 ms, TE = 60 ms) (STIR type with additional T2 contrast). (Figure courtesy of Sara Brockstedt, University of Lund, Sweden.)
Sievert (Sv)

(Radiation Protection) Sievert (Sv) is the unit given to the radiation protection quantities equivalent dose, and effective dose, as defined by the International Commission on Radiological Protection (ICRP 2007). It is also used for derived dose quantities such as committed equivalent dose and committed effective dose, and for ‘operational quantities’ defined by the International Commission on Radiation Units and Measurements (ICRU 1998) such as ambient dose equivalent and directional dose equivalent.

In SI units, 1 Sv is equivalent to 1 J of energy from incident ionising radiation absorbed in each kilogram of human tissues, that is:

\[ 1 \text{ Sv} = 1 \text{ J kg}^{-1} \]

In base SI Units, this can be more properly defined as

\[ 1 \text{ Sv} = 1 \text{ m}^2 \text{ s}^{-2} \]

where
- m is metres
- s is seconds

Sievert was named after the Swedish medical physicist Rolf Maximilian Sievert (1896–1966). For further information on use of the Sievert, see Related Articles detailed below.

Related Articles: International Commission for Radiological Protection, International Commission on Radiation Units and Measurements, Equivalent dose, Effective dose, Committed equivalent dose, Committed effective dose


Sigmoid dose–response curve

(Radiotherapy) The relationship between dose and radiation effect, i.e. the dose–response curve, is shown in Figure S.36. In general, it has a sigmoid (S) shape with the probability of effect tending to zero as the dose tends to zero and tending to 100% at very large doses. This applies to both tumour control and normal tissue complications, although the curve is often steeper for normal tissue damage than for tumour control such as in the case of a large inhomogeneous tumour containing sub-volumes with considerably different radiation sensitivities (e.g. hypoxic areas).

In the case of tumour control, the sigmoid shape can be explained from the random nature of cell killing after irradiation and the need to kill every cell (see article on Tumour control probability). However, for most normal tissue end-points, the biological interpretation of the sigmoid shape of the relationship is not obvious (see article on Normal tissue complication probability). Some authors have suggested a hypothetical tissue rescue unit (TRU), arguing that tissue breakdown occurs when the number of TRUs falls below a critical level. A TRU is defined as the minimum number of functional sub-units (FSU) required to maintain tissue function (see articles on Parallel organs and Serial organs for more on FSUs). However, this explanation for the sigmoid shape of the curve is questionable.

Many authors have proposed mathematical models to describe dose response and most have used one of three mathematical forms: the Poisson, the logistic or the probit model. The position of the dose–response curve is usually quantified by the dose required to obtain a specified level of response. The most frequently used position parameter is the \( D_{50} \), i.e. the radiation dose for 50% response. In the case of normal tissue complications, the \( D_{50} \), that is the dose producing a 5% incidence of complications, is often quoted since this is typically the acceptable incidence level for complications in patients receiving radiation therapy. Various measures have been used to quantify the steepness of the curve but the most commonly used is the normalised dose response gradient, denoted by \( \gamma \), introduced by Brahme in 1984. In this formalism, \( \gamma \) represents the increase in response, in percentage points, for a 1% increase in dose and its definition is given in Equation S.8. Thus \( \gamma \) is the product of slope and dose and is a dimensionless quantity.

![Sigmoid dose–response curve](Image 55x568 to 296x724)

**Figure S.35** Theoretically calculated beam profile of a circular transducer. The plane shown includes the central axis of the circular aperture. The aperture is actually located a small distance to the left of the figure for computational reasons. The aperture width is approximately four wavelengths. The main lobe is directed to the right, and side lobes can be seen above and below the main lobe at increasing angles.

angle, destructive interference causes a minimum. At increasing angles to the main lobe, alternate maxima and minima are formed, as can be seen in Figure S.35. The regions containing these maxima are referred to as side lobes. In Figure S.35, three side lobes can be seen aside from the main lobe.

**Sievert (Sv)**

In SI units, 1 Sv is equivalent to 1 J of energy from incident ionising radiation absorbed in each kilogram of human tissues, that is:

\[ 1 \text{ Sv} = 1 \text{ J kg}^{-1} \]

In base SI Units, this can be more properly defined as

\[ 1 \text{ Sv} = 1 \text{ m}^2 \text{ s}^{-2} \]

where
- m is metres
- s is seconds

Sievert was named after the Swedish medical physicist Rolf Maximilian Sievert (1896–1966). For further information on use of the Sievert, see Related Articles detailed below.

**Related Articles**: International Commission for Radiological Protection, International Commission on Radiation Units and Measurements, Equivalent dose, Effective dose, Committed equivalent dose, Committed effective dose


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![Sigmoid dose–response curve](Image 55x568 to 296x724)

**Figure S.36** The dose–response relationship is sigmoid in shape. This applies to both tumour control and normal-tissue damage.
The normalised dose-response gradient, \( \gamma \)

\[
\gamma = D \frac{\Delta P(D)}{\Delta D}
\]  

(S.8)

where \( P(D) \) is the probability of incidence with respect to dose \( D \).

This equation is useful in practical calculations where an estimate is required of the effect of a change in dose on the response of a tumour or normal tissue. However, since Equation S.8 corresponds to approximating the S-shaped dose–response curve to a straight line, the value of \( \gamma \) will depend on the response level at which it is evaluated. Clearly, evaluation at the bottom or top of the curve will produce a smaller increment in response for a given increase in dose than if it is evaluated on the steep part of the curve. Generally, the value of \( \gamma \) is written with an index indicating the response level at which it is defined, for example \( \gamma_{50} \) refers to the \( \gamma \)-value at a 50% response level.

For both tumours and normal tissues, the position and steepness of the dose–response curve will vary within a population due to their heterogeneous nature. They will also depend on the volume of tissue irradiated (dose–volume effect) and the fractionation scheme. Dose-response data are usually obtained from changing the total dose by one of two methods: either by increasing the dose delivered per fraction while maintaining the same number of fractions or by increasing the number of fractions while maintaining the same dose per fraction. However, for a specific end-point, the steepness of the dose response depends on which method is used. If the dose per fraction is increased, the dose–response curve is steeper than if the number of fractions is increased since this will result in an increase in the biological effectiveness per Gy.

**Abbreviations:** TRU = Tissue rescue unit, FSU = Functional sub-unit, \( D_{50} \) = Dose producing 50% response, \( D_s \) = Dose producing 5% response and \( \gamma_{50} \) = The normalised dose-response gradient defined at the 50% response level.

**Related Articles:** Dose response model, Fractions, Fractionation, Normal tissue control probability, Normal tissue dose response, Parallel organs, Serial organs, Therapeutic effect, Tumour control probability


**Signal aliasing**

(Nuclear Medicine) Signal aliasing occurs in MRI when an error in the spatial encoding of objects occurs. This is due to the anatomy being scanned exceeding the FOV in the phase encoding direction. Objects outside the FOV cannot be distinguished from objects inside the FOV and this leads to a spatial miscalculation of the area outside the FOV to the opposite side of the image. This causes problems when viewing the images as structures of interest are often obscured.

**Abbreviation:** FOV = Field of view.

**Related Article:** Wrap-around artefact

**Signal amplification technique (SAT) in PET**

(Nuclear Medicine) See Photomultiplier (PM) tubes

**Signal processor**

(Nuclear Medicine) Signal processing deals with the analysis and manipulation of signals. These signals can be of a variety of types: sound, images, electrical signals, etc. In nuclear medicine the signals of interest are invariably images.

Image processing techniques in nuclear medicine include image subtraction, spatial filtering, image reconstruction and the production of time activity curves.

These tasks are all performed by microprocessors. For very specific tasks an applications-specific integrated circuit (ASIC) is used. These devices are used in applications such as analogue-to-digital conversion and pulse height analysis.

Another type of microprocessor called a digital signal processor can be used to perform very fast real-time signal processing such as 3D image rotation.

**Signal-to-noise ratio (SNR)**

(Diagnostic Radiology) The signal-to-noise ratio of a detector is a measure of the detector’s performance. It is defined by the ratio of the total measured signal, \( S \), and the total system noise (standard deviation of the measured signal), \( \sigma \):

\[
\text{SNR} = \frac{S}{\sigma}
\]

In diagnostic radiography the signal is typically a measurement of x-rays which are transmitted through a patient and attenuated by various anatomical structures.

All radiographic images contain noise. In an ideal system this noise is due to the stochastic fluctuation of the x-ray quanta, \( \sigma \), which is described by Poisson statistics so that the maximum SNR available is

\[
\text{SNR} = \sqrt{\frac{N}{\sigma^2}} = \sqrt{N}
\]

where \( N \) is the number of x-ray quanta incident upon the detector. This relationship assumes that the detector is ideal and thus only quantum noise is present. In real systems there are many additional sources of noise such as structural or inherent detector noise which is due to the structure of the detector components such as the film granularity of screen-film systems or the density and dimensions of the phosphor screen used in indirect digital detectors.

The above description assumes that both the signal and noise are scalar quantities. Both signal and noise can be treated as spatial frequency dependent, in this case the signal is \( S(f) \) and the noise becomes the noise power spectrum \( \text{NPS}(f) \). For more complex assessment of system performance, detective quantum efficiency \( \text{DQE}(f) \) can be calculated using the modulation transfer function \( \text{MTF}(f) \), and the noise power spectrum \( \text{NPS}(f) \).

**Related Articles:** Rose model, Noise equivalent quanta (NEQ), Noise power spectrum (NPS), Detective quantum efficiency (DQE), Modulation transfer function (MTF)


**Signal-to-noise ratio (SNR)**

(Magnetic Resonance) Signal-to-noise ratio is used to describe the relative contributions to a detected signal of the true signal received from the coils and random background noise. This is used as an image quality parameter to ensure that the system is working well.
correctly. If the SNR is found to be reduced, this could indicate one of a number of problems with the system including system calibration, gain, coil tuning, RF shielding, coil loading and image processing.

Improving the SNR will generally improve the quality of the image. There are several common methods used to increase the SNR. The first is to average several measurements of the signal, with the expectation that random noise will be reduced. The SNR can also be improved by sampling larger volumes. This is achieved by increasing the field of view, which will increase the size of the pixels and therefore the signal in each, and by increasing the slice thickness, which will decrease problems caused by the partial volume effect. The amount of noise can be reduced by reducing the bandwidth and/or by applying a pre-processing filter. SNR will also be improved by increasing the strength of the magnetic field used.

Choosing the most appropriate coil will also affect the SNR. The receiver coil should encompass the whole anatomical area of interest to obtain the best SNR possible. Surface coils will provide the best SNR but can only be used to image structures close to the surface of the patient.

The SNR of a system is measured using a uniform phantom. An image is acquired and five small ROIs are drawn on the image. One is placed in the centre and the rest at 3, 6, 9 and 12 o’clock around the edge of the image. In each ROI the mean pixel value and the standard deviation are recorded:

\[
\text{SNR of the ROI} = \frac{\text{mean pixel value}}{\text{standard deviation}}
\]

The SNR over the whole image is the average of the five SNRs measured.

**Absolute SNR**: It is also possible to measure absolute signal-to-noise ratio to give SNR in Hz/10^{-2}mL^{-1} instead of as a relative ratio. The principle of the method is to create a single gradient along a small tube of pure water. This produces a simple projection from which a basic signal to noise for 1 mL of water in a 1 Hz bandwidth may be calculated.

**Abbreviations**: RF = Radiofrequency, ROI = Region of interest and SNR = Signal-to-noise ratio.

**Signal-to-noise ratio (SNR)**

(Ultrasound) The signal-to-noise ratio (SNR) in an ultrasound imaging system can be defined as the maximum instantaneous received signal power divided by the noise power. The received signal strength from a scatterer is dependent on attenuation of both the incident and the scattered sound wave, and consequently the SNR is depth dependent. The noise may arise from the transducer, preamplifier and A/D-converter. To optimise the SNR an optimal reception filter is chosen to be equal to the bandwidth of the receiver system (mainly the transducer). In other words this filter should match the bandwidth of the received signal, and is therefore called a matched filter.

The term SNR can also be used in connection with speckle statistics, where it is defined as the mean received signal amplitude divided by its standard deviation. Since the signal received from a large number of point scatterers is Rayleigh distributed, this relation is equal to

\[
\sqrt{\pi(4 - \pi)}
\]

as a consequence of the definition of mean and standard deviation for this distribution. This number approximately equals 1.91 or 5.6 dB.

---

**Silicon**

(General)

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Si</th>
</tr>
</thead>
<tbody>
<tr>
<td>Element category</td>
<td>Metalloid</td>
</tr>
<tr>
<td>Mass number A</td>
<td>28</td>
</tr>
<tr>
<td>Atomic number Z</td>
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</tr>
<tr>
<td>Atomic weight</td>
<td>28.085 kg/kg-atom</td>
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<td>Electronic configuration</td>
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</tr>
<tr>
<td>Boiling point</td>
<td>3538 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>2330 kg/m³</td>
</tr>
</tbody>
</table>

**History**: Silicon was first identified by Lavoisier in 1787 and first isolated in reasonably pure amorphous form by Berzelius in 1824. It is the second most abundant element by mass in the Earth’s crust after oxygen. It is found as silicon dioxide (silica) which occurs in many different mineral forms, and as silicates which are minerals containing silicon, oxygen and one or more metals. Its main use is as a major constituent of construction materials such as stone, concrete, cement and glass. Silicon is used as an alloying element, particularly in combination with aluminium to produce a castable alloy. Silicon is also the base material for the vast range of useful compounds with oxygen and/or carbon known as silicones. In recent times, the ability of silicon to act as the basis for semiconductors and computer chips has had a massive impact on modern life.

**Medical Applications**: Imaging panels – Silicon-charged particle detectors have good energy and spatial resolution and an excellent signal-to-noise ratio. Thus silicon is widely used in diagnostic imaging, most notably in amorphous silicon flat panel detectors.

Biomedical engineering – Silicon is used in the design of temporary therapeutic scaffolds (support structures) which assist or even promote the body’s natural healing processes. When constructed from silicon such scaffolds are biodegradable and do not require surgical removal.

Devices that rely on an electrical impulse to stimulate the body’s natural rehabilitative functions (e.g. defibrillators, pacemakers and cochlear implants) are silicon-based. Silicone rubber is found in mammary prostheses, finger joint prostheses and catheters.

**Related Article**: Imaging

**Silicon diode detector**

(Diagnostic Radiology) The silicon diode detector is an example of an indirect digital detector system used in medical imaging (x-ray). In its simplest form, it consists of a phosphor screen coupled to an amorphous silicon photodiode. The incoming x-rays interact with phosphor screen which absorbs the incident radiation, and converts it to visible wavelength light photons. The wavelength of the radiation detected is dependent on the material used for in the phosphor screen. The use of a scintillation crystal is what makes it an indirect method. Although detectors based upon silicon diode technology can be used to detect almost any wavelength of radiation, in diagnostic radiology the detectors are most used to detect x-ray wavelength radiation and a digital radiographic image is produced by the use of an array of pixels each formed by an individual diode, a pixel for which is shown in Figure S.37.

In x-ray image detection the scintillation crystal is typically made of caesium iodine most commonly doped with thallium. There are several reasons why this is used; firstly the k edges of CsI are at 33 and 36 keV which gives a high absorption probability...
of the x-ray energies generally used in radiography. The second advantage of using a CsI phosphor over the traditional phosphor material is that it can be grown in columnar crystals, which act like fibre optics. This allows a minimal level of lateral spread of light, and thus a high spatial resolution when coupled to photodiode pixels. This columnar structure also permits the application of relatively thick layers of CsI which increases the probability that an incoming x-ray photon will interact with the detector without loss of spatial resolution. Unlike the conventional powdered phosphors, spatial resolution is less limited by the diffusion of light.

These fluorescent light photons emitted by the phosphor screen are then converted into electric charge by a photodiode which stores the charge in the diode’s intrinsic capacitance. The photodiodes are constructed out of amorphous silicon, usually in the form of n-i-p diodes (although schottky and metal-insulator-semiconductor [MIS] structures are also used); the capacitance of the n-i-p structure can be described by the parallel plate formula with a \(100 \times 100 \mu m\) photodiode having a capacitance of \(\approx 1 \, pF\). The diodes are very thin (0.5–0.8 \(\mu m\)) and have high quantum efficiency.

To detect an image, many pixels form an active matrix array, with each pixel formed by a photodiode and a thin film transistor switch/gate, Figure S.30. To detect the image signal, the charge stored in each pixel is drained by the charge collector electrode and read out row by row. The gate line driver controls this readout process by applying a switching voltage to the transistor gates in each row.

An important aspect in pixel structure and detector sensitivity is the fill factor of each pixel, Equation S.7. This is due to the placement of the TFT within the pixel as it uses some of the active detection area that the photodiode cannot utilise and thus any radiation incident on this area will not be detected:

\[
\text{Fill factor} = \frac{\text{Light sensitive area}}{\text{Full area of pixel}} \tag{S.9}
\]

Although some research is currently investing pixel arrangements with 100% fill factor, most current medical imaging silicon diode detectors have a fill factor of approximately 70%.

A modern example of an indirect x-ray system uses a caesium iodide scintillator doped with thallium (CsI:Tl) with an amorphous silicon photodiode matrix panel (which is often made of 4 single panels, what is technologically cheaper). The panel includes 3000 \(\times\) 3000 pixel array (each pixel being 143 \(\mu m\)). This way the total active area is 43 \(\times\) 43 cm, with 12-bit greyscale resolution (4096 levels) and a maximum detective quantum efficiency of approximately 70%.

**Abbreviations:** MIS = Metal-insulator-semiconductor and TFT = Thin film transistor.

**Related Articles:** Indirect digital radiography, Storage phosphor, Dark current, Flat panel detector, Active matrix array, Detector quantum efficiency (DQE)


**Silicon diode detector**

*Radiotherapy* The irradiation of a semiconductor material can result in the creation of a hole-electron pair, provided enough energy is given to the electron to raise it into the conduction band and therefore a semiconductor detector is the solid state analogue of an ionisation chamber. Semiconductors can be used in the form of a silicon diode which is a p-n junction diode. The dosimeters are produced by taking n type or p type silicon and counter-doping the surface to produce the opposite type material. These diodes are referred to as n-Si or p-Si dosimeters, depending upon the base material. N-type or p-type diodes behave differently because their minority carriers are holes or electrons, respectively.

Due to its high density the sensitivity of a diode detector exceeds that of similar gaseous detectors by a factor of several tens of thousands, which implies that a point-like detector can be designed with a sensitive volume of less than 1 \(mm^3\). In the boundary between two regions one of p-type and another of n-type silicon there is a depletion layer which is free of charge carriers. When the detector is operating with zero external voltage, a potential difference of about 0.7 V exists over this depletion area, causing the charge carriers created by the radiation to be swept away into the body of the crystal. The sensitivity of the detector depends on the lifetime of the charge carriers and consequently on the amount of recombination centres in the crystal, which is determined by the diode type, the doping level and the accumulated dose. The diode sensitivity decreases with the accumulated dose as the radiation induces recombination centres into the lattice. The effect of radiation damage represents the main limitation of silicon diodes. Other effects related to the detector material have also to be considered. The output signal of the diode depends on the photon energy because of the higher atomic number of silicon (\(Z = 14\)) compared to soft tissue (\(Z \approx 7\)) and the resultant higher contribution to the signal from the photoelectric effect. The diode signal is also dose rate dependent because at high dose rates the recombination centres will be occupied resulting in a relatively lower rate of recombination. Moreover the radiation damage may change the dose rate dependence of diode.

An increase of the diode response with the temperature has been reported as increasing the detector temperature the energy of the minority carriers will increase and their probability for escaping recombination also increases. The thickness and the shape of the build-up cap will influence the angular response of the diode. Diodes can be used both for in vivo measurements and those inside phantoms. Diodes are particularly useful for measurements in high-dose gradient areas such as the penumbra region and in small field.
dosimetry. They are also often used for measurements of depth doses in electron beams measuring directly the dose distribution in contrast to the ionisation distribution measured by ionisation chambers. Diodes are also widely used in routine in vivo dosimetry. To determine the diode calibration factor for in vivo dosimetry a set of correction factors has to be established to account for variations in diode response in situations different from the reference condition.

**Silicon-controlled rectifiers (SCRs)**

*(General)* Silicon-controlled rectifier (SCR) is a type of thyristor. This four-layer, three-terminal solid state rectifier is controlled by gate signal.

The SCR is made up of four layers of semiconductor material arranged PNPN. The function of the SCR is similar to the diode, but its operation is best explained in terms of transistors. The anode is attached to the upper P-layer; the cathode, C, is part of the lower N-layer; and the gate terminal, G, goes to the P-layer of the NPN triode (Figure S.38).

SCR’s main application is as a fast switch that can turn on or off any amount of power without involving any moving parts. The SCR can often replace much slower and larger mechanical switches.

**Silver**

*(General)*

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Ag</th>
</tr>
</thead>
<tbody>
<tr>
<td>Element category</td>
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<tr>
<td>Mass number A</td>
<td>108</td>
</tr>
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<td>Atomic number Z</td>
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<tr>
<td>Atomic weight</td>
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<tr>
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</tr>
<tr>
<td>Melting point</td>
<td>1,235 K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>2,435 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>10,500 kg m⁻³</td>
</tr>
</tbody>
</table>

**History:** Silver has been known since ancient times because, like gold, it sometimes occurs as a native metal. It is commercially extracted from ores containing various combinations of copper, nickel, lead and zinc. Peru and Mexico have been producing silver since the sixteenth century and are still the two top silver-producing nations. In addition to its use as a precious metal, silver has many applications in the electrical industry, photography, optics and medicine.

**Medical Applications:** X-ray film emulsion – Silver’s main application in medicine is as a component of x-ray film emulsion. X-ray films are similar to those used for normal photography, except that the emulsion (containing silver bromide attached to a gelatin base) is about ten times thicker. The thicker emulsion increases the probability of interaction between the x-rays and silver bromide.

The interaction process within film can be summarised as follows: an incident x-ray is absorbed by a silver bromide crystal (usually via the photoelectric process) and its energy is transferred to an electron. This electron goes on to produce other free electrons (by ionisation) which can become trapped in faults in the crystal lattice referred to as ‘sensitivity specks’ or ‘sensitivity centres’. The trapped electrons attract positive silver ions and separate them from the bromine ions. The bromine atoms escape into the gelatin, while the silver atoms are left behind, forming a latent (invisible) image of trapped silver atoms. Chemical processes can then be applied to the film to create a visible image.

**Related Articles:** Film emulsion, Emulsion layer, Latent image

**Silver bromide** *(Diagnostic Radiology)* Silver bromide (AgBr) is a photosensitive compound that is the typical active element in a photographic/x-ray film (the photographic emulsion).

A brief explanation of the AgBr photographic image formation is as follows:

- When a light (or x-ray) photon excites a bromine atom it loses an electron.
- These free electrons are trapped into crystal defects of the photographic emulsion.
- The (+) silver ions are attracted into these negative defects, where they are neutralised and become Ag atoms (sensitised grains).
- The combination of areas in the film with different number of sensitised grains forms a LATENT IMAGE.
- During the process of film development the sensitised grains are stabilised (the exposed AgBr crystals being reduced to stable Ag atoms).
- During the next process of film fixing the remaining unsensitised grains (which had not been exposed to light photons) are removed and washed out.
- The final visible image contains areas with various darkness (optical density) depending on the concentration of Ag atoms, which in turn is proportional to the intensity of light or x-ray beam.

**Simulated annealing algorithm** *(Radiotherapy)* Simulated annealing is an optimisation technique used for inverse radiotherapy treatment planning.

In inverse planning a map of the desired distribution is specified. This may be the dose, a function of dose such as tumour control probability (TCP), dose volume histogram, or dose limits. The dose is then calculated for the desired beam orientations (in external beam radiotherapy) or source positions (in brachytherapy) and a cost function (C) constructed describing the difference between the prescribed (Dpres) and calculated (Dcalc) dose distributions:

\[
C = \sum_{x,y,z} (D_{\text{pres}} - D_{\text{calc}})^2
\]

x, y, z describes summation over 3D space. The weights (i.e. intensities) of the beams or sources are then adjusted iteratively to minimise this cost function.

The simulated annealing aspect is in the adjustment of the beam weights. A beam element is chosen at random and intensity added or subtracted from it. If it reduces the cost function then the change is kept. If it increases the cost function then it is kept with a probability, P, depending on the size of that change, ΔC, and a parameter called the temperature, T, via the equation:

\[
P = \exp(-\Delta C/kT)
\]

where k is a constant. The temperature is decreased as the optimisation progresses. Allowing changes which increase C allows...
the optimisation to escape from local minima in the cost function. Decreasing \( T \) with time ensures the optimisation settles to a solution. Simulated annealing has the advantage over most other inverse planning optimisation methods, such as gradient descent, in that it avoids trapping of the problem in local minima. Local minima may exist if the optimisation allows beam angles to vary or the cost function is expressed in terms of a non-linear function of dose, such as TCP.

**Abbreviation:** TCP = Tumour control probability.

**Related Articles:** Inverse radiotherapy planning, Interactive planning

**Simulator**

*(Radiotherapy)* Simulators provide the ability to reproduce the treatment set up and mimic most treatment configurations attainable on megavoltage treatment units. It is then possible to visualise the resulting treatment fields on radiographs or under fluoroscopic examination as though the observer were looking along the beam, that is ‘a beam’s eye view’. This view can be used to check the geometrical accuracy of the set up before treatment. Simple treatments such as parallel pairs can also be planned on a simulator when CT or other images are not required.

Simulators consist of a gantry and table arrangement as similar as possible to that found on isocentric megavoltage treatment units, with the exception that the radiation source in a simulator is a diagnostic quality x-ray tube rather than a high-energy linac or a cobalt source. Some simulators have a special attachment that allows them to collect patient cross-sectional information similarly to a CT scanner; the combination is referred to as a simulator CT. Figure S.39 shows a conventional treatment simulator. The photons produced by the x-ray tube are in the kilovoltage range and are preferentially attenuated by higher Z materials such as bone through photoelectric interactions. The result is a high-quality diagnostic radiograph with limited soft tissue contrast but with excellent visualisation of bony landmarks and high Z contrast agents. A fluoroscopic imaging system may also be included and would be used from a remote console to view the patient’s anatomy and to modify beam placement in real time.

**FIGURE S.39** A conventional treatment simulator has the capability to reproduce most treatment geometries available on radiotherapy treatment units. Simulators use a diagnostic x-ray tube and fluoroscopic system to image the patient.

**Sinc function**

*(General)* The sinc \( (\text{sine cardinal, sinus cardinalis}) \) function can be seen in Figure S.42. There are two commonly used versions of the function: the *normalised* sinc function and the *unnormalised* sinc function.

The sinc-normalised function is defined as follows and is plotted with the dotted line in Figure S.42. The function crosses zero at integer values. This version is commonly used in signal processing:

**FIGURE S.40** A modern simulator unit has the capability to screen the patient in fluoroscopic mode and produce digital images of the treatment area with field margins and multi-leaf collimator shapes and other shielding.

**FIGURE S.41** A typical simulator radiograph for a head and neck patient. The field limits and shielding are clearly indicated on the radiograph.

Figures S.39 through S.41 show a simulator, the console monitors and a radiograph showing the treatment field and areas of shielding.

**Sinc function**

*(General)* The sinc \( (\text{sine cardinal, sinus cardinalis}) \) function can be seen in Figure S.42. There are two commonly used versions of the function: the *normalised* sinc function and the *unnormalised* sinc function.

The sinc-normalised function is defined as follows and is plotted with the dotted line in Figure S.42. The function crosses zero at integer values. This version is commonly used in signal processing:
The sinc unnormalised function is defined as follows and is plotted with the solid line in Figure S.42. The function crossed zero at multiples of pi. This is historically used in mathematics:

\[
\text{sinc}(x) = \frac{\sin(\pi x)}{\pi x}
\]

For both forms, the value of sinc at \( x = 0 \) is unity.

**Sinc filter:** The Fourier transform of the sinc function is a rectangular function ('top-hat' function). Hence the application of a sinc filter in the time domain, would theoretically give the 'ideal' low-pass filter, eliminating all frequencies above the cut-off frequency, while leaving those below the cut-off frequency preserved.

However, implementation of the ideal sinc filter is impossible, since the function has infinite extent. Hence it is often truncated for real signals – the windowed sinc filter.

The sinc function is also used commonly as RF pulse waveforms in MRI for obtaining sharp slice profiles.

**Related Article:** Fourier transform

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**Single-channel analysers (SCAs)**

(Radiation Protection) A single-channel analyzer is a pulse height analyzer used for only a single energy. The selection of the pulses height (\( V \)) intervals, called channels, is achieved by energy discriminators: lower level (LLD) and upper level (ULD) or the baseline (\( E \)) and window (\( \Delta E \)).

For more information see Pulse-height analyzers (PHAs) for radiation detectors.

**Related Articles:** Proportional counter, Pulse-height analyzers (PHAs) for radiation detectors, Scintillation detector

**Single element transducer**

(Ultrasound) Single element transducers can be used for A-mode, B-mode and pulsed Doppler imaging. Therapeutic ultrasound devices use single element transducers.

For diagnostic ultrasound, the single element is used for transmit and receive. The pulse echo technique is used in:

**A-mode scanning:** The echo back shows the range of amplitudes from the beam direction. An A-line transducer is shown in Figure S.43.

**B-mode imaging:** The single element has to be swept or rotated about an axis to provide the sweep of echoes to reconstruct the B-mode. These mechanical scanners were the first used for B-mode imaging and can still be found today but are no longer used for high-end scanners where array transducers now dominate.

**Pulsed Doppler:** Single element transducers can be used for pulsed Doppler. An intravascular 20 MHz transducer is shown in Figure S.44.

**Related Articles:** Mechanical transducers, A-mode scanning, Pulsed wave Doppler

**Single exposure**

(Diagnostic Radiology) The most often used radiographic operation mode of an x-ray equipment. It produces only one x-ray exposure per examination (e.g. one chest radiograph) using the preset parameters (kV, mA, ms, or mA s). This mode differs from the multiple exposure mode, used most often in x-ray angiography, where a pre-planned sequence of x-ray exposures is performed during the examination.
Single phase generator

(Diagnostic Radiology) Less powerful x-ray equipment with classical high voltage generator use single-phase mains supply. Most often this type of generator is used for dental and mobile x-ray equipment. This high voltage generator uses single phase high voltage transformer. The rectifiers deliver mainly two-pulse kV waveform (100% pulsations). See typical circuit of this generator in the article High-voltage generator.

Usually most contemporary medium frequency generators use also single-phase mains supply, but the term single phase generator is used only for the older classical generators.

Related Articles: High-voltage generator, Voltage waveform

Single phase transformer

(Diagnostic Radiology) See Transformer

Single photon emission computed tomography (SPECT)

(Nuclear Medicine) Single photon emission computed tomography (SPECT) is a tomographic imaging method/system. A SPECT system typically consists of one or two scintillation cameras. The scintillation cameras will measure the distribution of administered activity from different angles to obtain projections. The scintillation cameras are mounted on a rotating gantry so that the cameras can be rotated around the patient in order to attain all projections. Acquiring two projections simultaneously will speed up the acquisition by a factor of two or more. This is of great use in dynamic studies.

The 2D image projections acquired are used to reconstruct a 3D image volume. There are a number of different algorithms used for image reconstruction which are described in separate articles.

The spatial resolution in the resulting image depends on the source-to-detector distance. Scintillation cameras in newer SPECT systems are designed to orbit close to the patient in order to minimise the source-to-detector distance. The cameras can follow the contour of the patient which will result in an improved spatial resolution.

SPECT is most frequently used to study myocardial perfusion. The studies provide information about a possible coronary artery disease or myocardial damage following infarction. The myocardial perfusion study is performed twice, once while the patient is resting and once while the patient exercises to see differences in cardiac perfusion. Sometimes the SPECT system is gated to map a certain part of the cardiac cycle. This is usually done with an electrocardiogram.

Another area in which SPECT is commonly used is cerebral perfusion. Important information about diseases like Alzheimer’s, seizure disorders and cerebrovascular diseases is gained by using SPECT.

Another application is in oncology. Some radiotracers accumulate in cancer cells. SPECT images will therefore show an increase in uptake in regions with a high concentration of cancer cells. This technique can find metastases at an early stage and make the treatment more effective. Displayed in Figure S.45 are anterior (Figure S.45a) and posterior (Figure S.45b) views of two bone scans. The patient in Figure S.45a demonstrates no pathological uptake of the radiopharmaceutical. The patient in Figure S.45b shows an increased uptake of radioactive tracer in metastases originating from a prostate cancer.

The SPECT system can be combined with other imaging techniques such as computed tomography (CT) and magnetic resonance tomography (MRT). This combined image system can provide both functional and morphological information.

Consider trying to locate metastases using only images provided by SPECT. The resulting images could prove occurrence of metastases but not yield any precise information about the metastases location. Using a combined image system the morphologic and functional images can be matched to provide both functional and positional information. These combined images can help increase the accuracy of the diagnosis and treatment. The CT images can also be used in the SPECT reconstruction to correct for attenuation.

Related Article: Scintillation camera


Single tank generator

(Diagnostic Radiology) See Mono block generator

Single voxel spectroscopy

(Magnetic Resonance) Single voxel spectroscopy (SVS) is the term used to describe a class of magnetic resonance spectroscopy (MRS) techniques in which signal is acquired from a localised volume (usually of cuboidal shape).

Localised signal acquisition is an important prerequisite for in vivo MRS, since otherwise the spatial extent of the region from which signal is acquired is limited only by the sensitivity of the RF coil. SVS has for some years been the most widespread approach to this, although it has now been eclipsed to some extent by chemical shift imaging (CSI).

The localisation pulse sequence typically consists of number of pulses applied along each of the three Cartesian axes in turn, so that the intersection of the three selected slices selects a volume of interest (VOI). Depending on the specific technique, signal may be acquired from this region directly (e.g. STEAM, PRESS), or post-processing may be required (chiefly ISIS and its variants).

The VOI selected using SVS is often only an approximation to the shape of most anatomical structures (although in some techniques, slices may be made oblique or a more elaborate pattern of intersecting slices may be employed to sculpt the selected volume more precisely to the anatomical region

![Image](a) (b)
of interest). However, the ability to position and size the VOI freely by manipulating the RF and gradient pulses was a major advance over previous techniques when SVS was first developed in the 1980s. In commercial implementations, the user is able to prescribe the VOI graphically at the MR scanner console (Figure S.46).

The aim of SVS is to collect as much signal as possible from within the VOI, while eliminating signal from outside this region entirely. Unfortunately, there is inevitably some signal loss and a degree of contamination with extraneous signal. Protocols and phantoms have been developed to assess the performance of SVS techniques in this regard.

An important issue with most SVS techniques is that the use of frequency selectivity to achieve slice selection means that each resonance peak within the spectrum originates from a slightly different spatial location. This chemical shift offset is minimised by using high switched gradient amplitude.

Related Articles: Chemical shift imaging, ISIS, PRESS, STEAM, Magnetic resonance spectroscopy, MRS voxel contamination


Sinogram

(Nuclear Medicine) In SPECT and PET, images are acquired from different angles and they produce a 2D projection of the radionuclide distribution. One can assume that a single trans-axial row in such a projection would be 1D representation of the activity distribution in that cross section. A 1D distribution from all projections is used to reconstruct a cross-section image. A common way to represent these distributions is in a sinogram, which is a 2D matrix where the successive rows represent successive projection angles. If a source is located off centre the source traces a sinusoidal track along the matrix; hence the name sinogram. A sinogram is a convenient way to represent data collected from different projection angles and sinograms are often used to determine the cause of image artefacts. See article CT reconstruction.


Skin dose

(Radiation Protection) Skin dose is the absorbed dose, in Gray (Gy), incurred by an irradiated area of skin. Skin dose is calculated by using exposure factors (e.g. x-ray kVp and dose-area-product).

Of particular concern with skin irradiation is the risk of causing erythema. There is a threshold for causing erythema of about 5 Gy, and effects become progressively worse with higher dose. Skin dose is calculated additively for all exposures in a single procedure. However, effects are localised to the area of exposure, so an individual may incur separated exposures that contribute to a total skin dose in excess of 5 Gy without actually causing erythema because the total dose was not delivered to the same area of skin.

Skin dose calculations therefore also need to take into account the area of exposure in order to assess the risk of erythema. The skin dose limit defined by ICRP is an equivalent dose of 500 mSv averaged over an area of no greater than 1 cm². Repeated exposures to the same area will combine additively towards the erythema risk, while exposures to different areas will not (although all exposures contribute towards a calculation of overall skin dose).

In radiotherapy skin dose is one of the limiting factors affecting the treatment plan. Rotational geometries spread the exposure over circumference of the patient, reducing the erythema risk (and dose to other radiosensitive organs near the tumour site), while still delivering the prescribed target dose to the tumour.

Related Articles: Equivalent dose, Deterministic effects


Skin reference marks

(Radiotherapy) In external beam radiotherapy treatment it is important to relate the internal treatment target to the treatment room and hence the delivery system. A key component of this is to mark the patient’s skin with reference marks. These are usually tattoos and are aligned to the treatment room using localisation lasers.

The first stage of the treatment process involves imaging the patient, for example with CT. The skin reference marks are often drawn at this point. At treatment simulation and each treatment visit, the skin reference marks are used to set the patient up.

Related Articles: Localisation lasers, Treatment verification, Radiotherapy

Skin sparing

(Radiotherapy) One of the advantages of megavoltage photon radiotherapy is the ability to treat deep-seated malignancies with relative skin sparing. This is a consequence of the low surface dose from the relatively long range of the secondary electrons, which carry energy deeper into the medium. Typical surface doses are ~20% for a MV photon beam, (Figure S.47) and increases with increasing field size and decreasing energy of beam. Orthovoltage/superficial photon beams do not exhibit skin sparing, and hence are used for shallow lesions where a high skin dose is beneficial. In cases where a thermoplastic shell is used for immobilisation, it is...
common for the shell to be cut out around critical structures so that the full effect of the skin sparing can be utilised.

MeV electron beams display a limited amount of skin sparing, with the surface dose typically 80%. In the case where skin sparing is not desired, to obtain a more uniform depth dose distribution within the tissue, wax bolus can be placed on the skin as tissue substitute.

**Related Article:** Percentage depth dose


**Slant-hole collimator**

*(Nuclear Medicine)* A slant-hole collimator is basically a parallel-hole collimator with angled holes, typically 25°. Due to the design the collimator can be positioned closer to the patient in some patient studies, for example left anterior oblique cardiac views. When the source to collimator distance decreases the spatial resolution increases, which is why it is desirable to get as close to the patient’s body as possible. The collimator characteristic is similar to those of the parallel-hole collimator (see separate article).

**Related Articles:** SPECT, Collimator, Parallel-hole collimator, Diverging collimator, Converging collimator, Collimator design, Collimator parameters


**Slew rate**

*(Magnetic Resonance)* In MRI, the term *slew rate* denotes the ratio between a magnetic field gradient (often measured in mT/m, millitesla per meter) and the ramp time (see also *Ramp time*), for example the time it takes for the gradient to build up by driving a current through one or more of the gradient coils (see *Magnetic field gradients*).

The parameters minimum ramp time and maximum gradient strength are governed by health aspects, since when a magnetic field variation $dB/dr$ is built up or turned off a temporally varying magnetic field $dB/dt$ interacts with the body according to

$$\frac{dB}{dt} = \frac{(dB/dr) \times L_s}{2/\Delta t} = \frac{SR \times L_s}{2} \quad (S.10)$$

In *Equation S.10*, $dB/dt$ denotes the magnetic field variation over time, $dB/dr$ is the intended gradient strength in the $r$ direction after ramping up, $L_s$ is the effective length of the gradient coil in the $r$ direction and $\Delta t$ is the ramp time for the gradient to reach the value $dB/dr$. The slew rate (SR) is, with the notation according to *Equation S.10*, defined as the ratio between $dB/dr$ and $\Delta t$ and hence a direct multiplication between SR and $L_s/2$ gives the biologically important parameter $dB/dt$ at position $L_s/2$ (Figure S.48).

In modern MRI scanners, minimum ramp times are of the order of a few hundreds of microseconds. Using *Equation S.10* above, SR can be calculated for any gradient applied. As an example, consider a gradient coil with an effective length of 0.5 m in the $z$ direction, a desired gradient in the same direction of $dB/dz = 30$mT/m and a ramp time of 300μs. Insertion of these values in *Equation S.10* gives:

$$SR = \frac{(dB/dr)/\Delta t}{30 \text{ mT/m}/0.3 \text{ ms} = 100 \text{ (T/m)/s}} \quad (S.11)$$

**Related Articles:** Ramp time, Gradients

**Slice position**

*(Magnetic Resonance)* Errors in slice position can be very important in a clinical context. If measurements from MR images are to be used in radiotherapy or surgical planning, then slice position accuracy is vital.

The main factors which affect slice position accuracy are

1. Calibration of the slice select gradient
2. Frequency settings
3. Misadjustment of the measurement system on the scanning couch

**Measurement of Slice Position:** The standard test object used to measure slice position accuracy is EuroSpin Test Object 3: a Perspex cylinder filled with copper sulphate solution and 16 pairs of crossed rods (MR cold targets). Each pair of rods intersects midway through the test object in the axial direction.

The simplest test to perform is a single slice calibration test. Here the centre point of TO3 is accurately aligned with the light

**Further Reading:**
beams, such that a slice midway through the test object can be acquired. For perfect slice position, the rod separation in the resulting image would be zero. Any increased rod separation would be indicative of slice position error.


Slice profile

(Diagnostic Radiology) In computed tomography, the slice profile is also referred to as the slice sensitivity profile, z-sensitivity profile or slice width profile. It is a measure of the sensitivity of a detector to objects at different positions along the scan axis (z-axis), and is obtained by plotting of CT number against z-axis position (Figure S.49). The full width at half maximum of the slice sensitivity profile defines the slice thickness.

The slice sensitivity profile can be measured using a thin angled plate in axial (sequential) or a thin disc in helical scanning Edyvean et al. (2003) – see a similar example in Figure S.52.

Axial (sequential) scanning results in the most ideal slice sensitivity profile. In helical scanning, the slice sensitivity profile broadens if the pitch is increased (Figure S.50).

Related Article: Slice thickness


Slice profile

(Magnetic Resonance) A slice profile is a plot of signal intensity along a line that perpendicularly bisects a slice (Figure S.51).

The ideal slice profile would be a top hat function with constant signal intensity over the slice width and zero signal from outside the slice of interest. In practice, slice profiles are imperfect, typically being more rounded (as in Figure S.51).

Problems Associated with Poor Slice Profiles: Clinically, slices with poor profiles are undesirable: if a small region of pathology were to coincide with the edge of such a slice it could be hard to detect. Furthermore in multi-slice acquisitions with small slice spacing, poor slice profiles can lead to interference between adjacent slices (additional saturation and cross talk), which can reduce image SNR and uniformity. Two methods to reduce such interference can be adopted: firstly, the slice spacing can be increased (in which case small pathologies within the inter-slice gap may be missed); secondly, slices can be acquired in a non-consecutive (interleaved) order, for example odd slices and then even slices.

Factors Affecting Slice Profile: RF pulse envelope: ideally the RF pulse would have a sinc envelope extending infinitely throughout time – this would Fourier transform to give the perfect (top hat) slice profile. In practice, the sinc has to be truncated in time, giving a rounded slice profile. The more truncated the sinc, the worse the slice profile:

1. Gradient linearity and gradient gain
2. Main magnetic field inhomogeneity
3. TR/T1 ratio

Measuring Slice Profile: Typically test objects containing angled plates/wedges are used to measure slice profile (see Figure S.52).
Slice selection

Slice selection is one of the principal methods used for spatial localisation in MRI. It is used to restrict spin excitation to one- or more two-dimensional slices within the object or person being imaged. Other techniques are then used to localise signal within the selected slice.

Slice selection is normally achieved using a frequency-selective RF pulse in combination with a static field gradient. This approach is based on the resonant frequency of a nucleus being proportional to the strength of the magnetic field in which it is situated. Thus application of a linear magnetic field gradient results in linear variation in frequency along the direction of field variation (Figure S.53). An RF pulse with bandwidth \( \delta \omega \) can then be applied in the presence of the gradient to nutate magnetisation into the transverse plane only within a predetermined slice of material of thickness \( \delta z \), within which the resonant frequencies of spins correspond to the frequency content of the pulse.

It is desirable to select a slice with a rectangular profile, so that there is uniform excitation of material within the desired slice and none outside. This would require application of a sinc-shaped RF pulse, since a sinc function is the Fourier transform of a rectangular, sinc-like function, since a sinc function is of infinite extent, a compromise is necessary and it is usual to use truncated sinc pulses with Gaussian smoothing or, increasingly, numerically tailored ‘designer’ pulses. The aim is to achieve a slice profile as close to rectangular as possible within the pulse length constraints imposed by the remainder of the pulse sequence.

Because nutation occurs over the duration of the RF pulse, the resulting transverse magnetisation is dephased by the slice selection gradient. A rephasing gradient is applied immediately afterwards to compensate for this, although perfect rephasing is not possible. The properties of this additional gradient are determined empirically or by modelling, but its area (amplitude \( \times \) duration) is usually approximately half that of the slice selection gradient.

In a spin echo sequence, the refocusing pulse is usually also made slice selective, but no rephasing gradient is needed as long as the pulse and gradient are arranged symmetrically (Figure S.54).

Further Reading: Lerski, R. et al. 1998. Quality control in magnetic resonance imaging, IPEM Report 80, Institute of Physics and Engineering in Medicine (IPEM), York, UK.

Related Article: Slice thickness

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Related Article: Slice thickness

Slice sensitivity

(Diagnostic Radiology) See Slice thickness

Related Articles: Computed tomography, Multislice scanner, Helical pitch, Helical interpolation, Partial volume effect

Slice thickness

(Diagnostic Radiology) In computed tomography, the slice thickness is the thickness of the data volume imaged along the scan axis (z-axis). It is also referred to as the slice width, the slice sensitivity or the z-sensitivity. The slice thickness is defined as the full width at half maximum (FWHM) of the slice sensitivity profile (Figure S.49).

On single slice scanners the slice thickness is determined by the z-axis x-ray beam collimation. Slice thicknesses of between 1 and 10 mm can be achieved.

On multislice scanners the minimum slice thickness is determined by the data acquisition width, which is governed by the z-axis detector, or detector grouping dimension (see Multislice CT scanner). For the same data acquisition width different slice thicknesses can be reconstructed by altering the filter width used (see Helical interpolation).

Selection of slice thickness is determined by the clinical application. Narrow slices result in an improved z-axis spatial resolution.
and enable equal resolution in all planes, so-called ‘isotropic resolution’ (Figure S.55). Isotropic resolution enables high-quality multiplanar and 3-D reconstructions. Another advantage of narrow slices is a decrease in partial volume effect, so that contrast resolution is optimised (Figure S.56). Use of narrow slices also reduces partial volume artefacts which cause streaking in the CT image. However, one disadvantage of using narrow slice widths is that image noise (standard deviation of CT numbers) increases inversely with the square root of the slice thickness (Equation S.12). Halving the slice thickness therefore increases noise by a factor of 1.41:

\[ \sigma = \frac{1}{\sqrt{T}} \]  

(S.12)

where

- \( \sigma \) is the standard deviation of CT numbers or image noise
- \( T \) is the slice thickness

**Related Articles:** Computed tomography, Multislice scanner, Helical pitch, Helical interpolation, Partial volume effect

**Slice thickness**

(Magnetic Resonance) In MRI slice profiles may not be perfect (see Slice profile). For this reason the full width half maximum (FWHM) of the slice profile is used to define the slice thickness.

**FWHM:** The full width at half maximum (FWHM) is defined as the distance between the points where the intensity is half of the maximum value (see Figure S.57).

Clinically, the slice thickness used depends upon the anatomic region being imaged. Typically, a large organ such as the liver will be imaged with relatively thick slices (~7 mm), the brain will be imaged with medium slices (~5 mm), and small joints will be imaged with thin slices (~3 mm).

**Related Article:** Slice profile

**Further Reading:** Lerski, R. et al. 1998. Quality control in magnetic resonance imaging, IPEM Report 80, Institute of Physics and Engineering in Medicine, York, UK.
Slice thickness
(Ultrasound) The slice thickness of an ultrasound image is the distance in the elevation plane from which echoes arise. In most systems, the slice thickness varies with depth.

The impression of a 2D ultrasound image is that it is from a thin slice. However, in conventional transducers the thickness is governed by the width of the active elements and an acoustic lens (Figure S.58). The element/lens combination is chosen to optimise slice thickness at a depth appropriate to the frequency of the transducer and at which it will be used. A high-frequency transducer (e.g. 10 MHz) optimised for 1–2 cm depth has narrower elements than a transducer operating at 3 MHz optimised for 6–10 cm which requires a larger aperture in the elevation plane. For more superficial tissue, the 3 MHz transducer may have slice thickness of several mm.

Slice thickness is usually the worst of the three planes of spatial resolution (link) in ultrasound images but is often overlooked by users since, unlike axial (link) and lateral resolution (link), its effects are not always obvious in the displayed image. Poor control of slice thickness can adversely affect contrast resolution (link) as shown in Figure S.59. For objects smaller than the slice thickness, the ultrasound characteristics of the object are combined with adjacent tissue within the slice thickness at the same depth, so reducing contrast. Slice thickness can be improved by

**Annular arrays:** Mechanically swept annular arrays permit focussing in the transverse and elevation plane by using timing differences in transmission and reception. Annular arrays are no longer commonly available.

**2D arrays:** Rows of elements can be used to provide dynamic focussing in transmission and reception in the elevation plane.

**Variable thickness transducer elements:** Differences in transducer element thickness lead to differences in resonant frequency across the elevation plane with high frequencies used in the centre of the elements for superficial structures and low frequencies applied over a wider aperture for use at greater depths.

**Harmonic imaging:** It is postulated that because harmonics are generated in the highest intensity region of the beam, harmonic images inherently show echoes from the centre of the beam with a possible reduction in effective slice thickness.

Figure S.58 shows an image of slice thickness variation in a linear array. The width of the beam in the elevation plane is dependent on the transducer elements and the lens. In the diagram, the slice thickness (arrows) is optimised at depth A and is poor at depth B.

Figure S.59 shows the effect of variation in slice thickness in an ultrasound phantom. The tubes are 2 mm diameter and are imaged along their length by a 5 MHz linear array transducer. The diagram shows the end view of the transducer. At D, depth 2 cm the tube is clearly displayed. At larger depths, the slice thickness encompasses the tube and surrounding material (diagram). The image of the tube is less clear and the edges less clearly displayed.

**Slice warp** (Magnetic Resonance) Slice warp is defined as the shift of the midpoint of a slice in the slice select direction from its true orthogonal plane. This effect is rarely seen with modern-day scanners but can be caused by main field non-uniformity. The Eurospin test object TO3 can be used to look for slice warp. TO3 is made up of a series of pairs of rods perpendicular to each other spaced throughout the phantom. A variation in the separation of rod pairs over the image plane indicates a warping of the slice, with the degree of variation indicating the degree of warping present.

**Slip ring technology** (Diagnostic Radiology) Slip ring technology removes the need for cables to be connected to the rotating gantry and so enables continuous gantry rotation. Before the introduction of slip rings in CT design, the power from the generator to the tube was supplied by high-tension cables. Cables were also required to transfer power to other components, such as the collimators, and also to transmit the signal from the detectors. The gantry therefore had to stop and reverse direction after each rotation in order to unwind these cables. This resulted in long acquisition times and limited the use of CT in many clinical applications.

Slip rings consist of electrical conducting rings located on the stationary part of the gantry (Figures S.60 and S.61). The cables are connected to these stationary rings, while ‘brushes’ on the rotating
Sm-153-EDTMP [Lexidronam]

(Nuclear Medicine) Sm-153 Lexidronam or Quadramet® (CIS Bio International, France) is a therapeutic radiopharmaceutical for palliation of bone pain from metastatic prostate carcinoma.

**153Sm-lexidronam** consists of 153Sm labelled with ethylenediaminetetramethylene phosphonic acid (EDTMP). Quadramet is formulated as a sterile, non-pyrogenic, clear, and colourless to light amber isotonic solution of 153Sm-lexidronam for slow IV administration (infusion). The 153Sm-EDTMP activity recommended for therapy of bone pain is 37 MBq kg⁻¹ body weight. Thirty-five percent of the patients obtain pain relief after one week, rising to 70% within a month.

Samarium-153 is produced in high yield and purity by neutron irradiation of isotopically enriched samarium:

\[
^{152}\text{Sm}-\text{oxide} \rightarrow ^{152}\text{Sm}(n,\gamma)^{153}\text{Sm} \rightarrow ^{153}\text{Eu}.
\]

153Sm is a mixed beta emitter with max beta energies of 640 (30%), 710 (50%) and 810 keV (20%), and a gamma energy of 103 keV (29%). It has an average (CSDA) and maximum beta particle ranges in water of 0.5 and 3.0 mm, respectively. The half-life of 89Sr is 46.28 h.

After an IV injection the 153Sm-EDTMP is rapidly eliminated from the blood and avidly localises in reactive bone metastasis similar to 99Tcm-diphosphonates, but 153Sm undergoes a hydrolysis reaction at the bone surface. After 4–6 h 35% has been excreted by the urine. The bone uptake is about 65% of the administered activity and related to the extent of the metastasis.

The mean absorbed dose to bone metastasis from an IV injection of 153Sm-EDTMP has been estimated to be 33 mGy MBq⁻¹. The organ most exposed is not unexpectedly the bone surface and red bone marrow, which receive an absorbed dose of 6.8 and 1.5 mGy MBq⁻¹, respectively. The effective dose for 153Sm-lexidronam is approximately 0.31 mSv MBq⁻¹. Thus, a typical total activity of 2590 MBq, divided in several fractions, gives an average absorbed dose to the metastasis of approximately 86 Gy, 18 Gy to the bone surface and 4 Gy to the red bone marrow.


**Related Articles:** Sr-89-Chloride [Metastron], Re-186-HEDP

Small-animal SPECT imaging

(Nuclear Medicine) Imaging the internal distribution of radiopharmaceuticals in small animals using dedicated SPECT systems with high resolution. Different detectors and collimators are used. Most common is scintillation detectors and pinhole collimators. In some literature the instruments used are named micro-SPECT. This implies that the spatial resolution is submillimeter.

**Related Article:** Micro-SPECT


Snell’s law

(Ultrasound) Snell’s law describes the change in wave path direction due to refraction as it passes between different media where...
there is a change in wave velocity. Snell's law applies to light, sound and other wave phenomena, the law is named after the Dutch scientist Willebrord Snellius (1580–1626).

For ultrasound, the law describes the transmission angle when an ultrasound wave encounters tissue boundaries with differing propagation speed in the case of oblique incidence (Figure S.62). In the diagram, $\theta_i$ is the incident angle, $\theta_r$ the transmitted angle and $\theta_t$ the reflected angle. $\theta_t$ is equal to $\theta_r$, $c_1$ and $c_2$ are the speeds of sound in two tissue types:

$$c_1/c_2 = \sin \theta_r / \sin \theta_i$$

In ultrasound scanning, refraction leads to registration errors since the scanner assumes a straight path in transmission and reception of echoes. Refraction also leads to loss of signals and changes in contrast resolution as a result of variation in ultrasound intensity. In soft tissue there is generally little relative difference in speed of sound and refraction is not a severe problem. Refraction is more of a problem if ultrasound strikes bone at an oblique angle since there is a greater disparity between the speeds of sound. If ultrasound meets an interface at an angle exceeding the critical angle, then it is reflected from the interface and there is no onward transmission.

**Related Articles:** Speed of sound, Reflection coefficient

**SNR**

The term is an abbreviation for signal-to-noise ratio. For more information see the relevant articles in *Diagnostic radiology*, *Magnetic resonance imaging* and *Ultrasound*.

**Sodium**

(General)

<table>
<thead>
<tr>
<th>Property</th>
<th>Value</th>
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</thead>
<tbody>
<tr>
<td>Symbol</td>
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<tr>
<td>Element category</td>
<td>Alkali metal</td>
</tr>
<tr>
<td>Mass number $A$</td>
<td>23</td>
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<tr>
<td>Atomic number $Z$</td>
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<tr>
<td>Atomic weight</td>
<td>22.99 kg/kg-atom</td>
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<tr>
<td>Electronic configuration</td>
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</tr>
<tr>
<td>Boiling point</td>
<td>1156 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>970 kg/m$^3$</td>
</tr>
</tbody>
</table>

**History:** The compound sodium chloride, common salt, has been a valuable commodity since ancient times. The reactive nature of sodium results in the formation of very strongly bonded compounds, and it was not until 1807 that Humphry Davy used electricity to obtain sodium metal by electrolysis of molten sodium hydroxide. Elemental sodium is a soft metal which reacts violently with water, limiting its practical uses. It does find applications in sodium vapour lamps commonly used for street lighting, and liquid sodium has been employed as a coolant in nuclear reactors and as a reducing agent in the production of titanium. Compounds such as sodium carbonate and sodium hydroxide are employed in many industrial processes. Sodium ions are important in many physiological processes and sodium levels must be regulated at the cellular level within animals to maintain health.

**Medical Applications:** Scintillation counters – The very early workers with radioactivity were aware that some substances, such as zinc sulphide and diamond, are luminescent when exposed to x-rays. In modern times sodium iodide (activated by the introduction of small amounts of thallium) is commonly used as a luminescent scintillator in x-ray/γ detectors. Sodium iodide is ideal for this purpose as it is highly absorbing of x- and γ-rays, transparent (so that the flashes of scintillation light can escape to be counted) and it can be grown as large crystals.

**Related Articles:** Scintillator, Scintillation camera, Scintillation detector

**Software phantom**

(Nuclear Medicine) Software phantom is a term used for computer-made phantoms. These phantoms are used to gather information about the imaging system. The advantage of software phantoms is that the user knows (sets) all parameters and can adjust them in order to detect the influence of each parameter in a controlled environment. Software phantoms are used in Monte Carlo simulations. Monte Carlo programs can then be used to calculate radiation dose to a patient or specific organ when assuming a certain distribution and radiation behaviour (interactions probability distributions, organ attenuation coefficients, etc.).

An obvious disadvantage with software phantoms is the differences between the clinical situations and the simulated ones. Conclusions drawn from simulated data are not always valid on patients. For example, in the MIRD formalism, activity is assumed to be evenly distributed in a compartment. A compartment is typically an organ or part of an organ assumed to have the same size and location in every patient. It is also assumed that the attenuation coefficient is constant for the whole compartment. These assumptions are not true because the activation tends to accumulate unevenly over the compartment, patients are not all alike and the attenuation coefficients do change within the organ. Even though the limitations to the MIRD formalism are well known, it is still in use when calculating organ doses in nuclear medicine examinations. In the future more complex models with patient specific photons will be available as computer power increases.

**Related Articles:** Physical phantom, MIRD formalism

**Solenoid**

(General) A solenoid is a helical (spiral) coil of wire that produces a magnetic field when carrying electrical current (see also the article *Electro-magnet*). When a current is passed through the coil, the magnetic field within the coil is relatively uniform. In engineering, the term solenoid may also refer to a variety of transducer devices that convert magnetic energy into linear motion.

**Related Articles:** Electro-magnet, Stator
Solid state detectors

(Diagnostic Radiology) Solid state detectors use a crystalline semiconductor material (e.g., silicon, germanium) as a detecting medium. It consists of a p–n junction across which a pulse of current develops when a particle of ionizing radiation traverses it.

See Semiconductor detector.

Solid state rectifier

(General) Solid state rectifiers are a type of rectifiers based on solid state (semiconductor) diodes. The main characteristics of such a rectifier are low voltage drop, quick response, low cost.

See Rectifier

Solid water phantom

(Radiotherapy) The use of epoxy resin-based substances (aka solid water) as a substitute for water is common in radiation dosimetry. Solid water has radiation characteristics very close to those of water, and when used as a phantom for radiation in the radiotherapy energy range, any phantom-water corrections can be eliminated. The use of solid water phantoms is advantageous over traditional water phantoms, as they avoid the possibility of immersing chambers in water, reduce the reproducibility errors and are more convenient to use. Solid water is available in slabs of varying thicknesses that can be used to create specific depths of required build-up or to provide backscatter, as well as inserts to hold specific chambers, such as parallel plate chambers.


Solution

(General) Solutions are homogeneous mixtures that are composed of two or more substances. They contain a solute which is dissolved in a solvent. A solution is characterised by interactions between the solvent and solute which results in a reduction in free energy. An everyday example of a solution is that of sugar dissolved in water. It is also possible to dissolve a gas in a liquid, for example carbonated drinks in which carbon dioxide is dissolved in water. Furthermore the definition of solutions extends to solids, for example a metal alloy is described as a solid solution.

Concentration is a measure used to quantify the amount of solute dissolved in a solvent which can be expressed in parts per million (ppm) or molarity.

Related Article: Solvent

Solvent

(General) Solvents are fluids (liquids or gases) that can dissolve solutes in solid, liquid, or gaseous form, to produce a solution. The concept of solubility is used to quantify the amount of solute that is soluble in a given volume of solvent at a given temperature. There are two main types of solvent, polar and organic. A useful way to remember which type of solvent dissolves which type of solute is ‘like-dissolves-like’ that is a polar solute dissolves well in a polar solvent.

Polar: Polar solvents are made up of polar molecules. A molecule is described as polar if the atoms that make it up have differing electronegativities. The most common polar solvent in use is water.

Organic: Organic solvents are made up of molecules that contain carbon, they are non polar; common examples include acetone, ethanol and hexane. It should be noted that organic solvents are often highly flammable. Organic solvents are used in paint thinners, nail polish removers, perfumes and detergents.

Related Articles: Solution, Atom

Sonogram

(Ultrasound) Velocity data in pulsed Doppler measurements (or continuous-wave Doppler) is usually presented as a sonogram, where time is represented on the horizontal axis, and velocity (or Doppler shift) on the vertical. The brightness represents the received Doppler power at a given velocity. Figure S.63 shows a so-called triplex image where the anatomical greyscale image is shown with the velocity information in colour overlaid on the greyscale image. In the bottom part of the image is shown the sonogram. This gives a time-velocity representation of the velocity distribution within the range gate, which is indicated by the small lines perpendicular to the line indicating the insonation angle for the pulsed Doppler beam.

Sometimes a sonogram is also referred to as a spectrogram.

Sonography

(Ultrasound) Sonography describes the use of ultrasound, specifically medical diagnostic ultrasound. It is an abbreviation of ultrasonography. The term ultrasound describes sound frequencies which are inaudible to the human ear, that is above 20kHz. Typically the frequencies used range from 1 to 20MHz. An advantage of ultrasound over other diagnostic imaging modalities is that it is relatively inexpensive and is safe; it does not use ionising radiation and does not usually require contrast agents. An often quoted disadvantage is that it is user-dependent meaning the acquisition and interpretation of ultrasound images depends upon the training and skill of the operator.

The term sonographer is used to describe practitioners of medical ultrasound. The term sonologist has also been used to denote medical doctors using ultrasound but is less commonly used.

Sound attenuation

(Ultrasound) See Attenuation

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Related Articles:

- Solvent
- Rectifier
- Semiconductor detector
- Virtual Water™
- Solid state rectifier
- Solution
- Solvent
- Organic solvents
- Polar solvents
- Ultrasound
- Sonography
- Sonogram
Source axis distance (SAD)

(Radiotherapy) This describes the distance from the target source to the isocentre of the machine. Commonly it is 100 cm. It is widely used in radiotherapy for dosimetry and patient set-up. Linacs are isocentrically calibrated when the reference point is located at the reference depth at a distance from the source, equal to the SAD. Equivalently, cobalt units are normally calibrated ‘in-air’, with the reference point again located at the isocentre, equal to the SAD.

**Abbreviation:** SAD = Source axis distance.

**Related Articles:** Source axis distance (SAD), Source surface distance (SSD)

Source block distance (SBD)

(Radiotherapy) This describes the distance from the target source to the block tray, either to the upper side or in rare cases to the lower side as seen from the source.

**Related Articles:** Source axis distance (SAD), Source surface distance (SSD)

Source coordinates

(Nuclear Medicine) In nuclear medicine imaging this refers to the process by which a gamma ray entering the gamma camera is represented at the appropriate x-y position on the image.

When a scintillation event reaches the crystal, the light produced is distributed among a number of adjacent photomultiplier tubes (PMTs). The PMT closest to the event receives the maximum amount of light, so this represents its approximate position. A more accurate X–Y coordinate is obtained by considering the relative response of the other PMTs. The amount of light detected by a PMT is inversely related to the distance between its centre and the site of the interaction (Figure S.64).

In the older style analogue cameras, the position of an event is determined by splitting the signal from each PMT into four output lines known as X+, X-, Y+, Y-. The positioning voltages X and Y and the energy Z are determined using the following equations (K is a constant):

\[
Z = X^+ + X^- + Y^+ + Y^- \tag{S.13}
\]

\[
X = K(X^+ - X^-)/Z \tag{S.14}
\]

\[
Y = K(Y^+ - Y^-)/Z \tag{S.15}
\]

In the newer cameras, the output signal from each PMT is digitised. This is often analogous to the signals described above and the same equations are used. Because the positional information is determined using software, accuracy can be improved by incorporating weighting factors which take into account any non-linearity in PMT response.

**Related Articles:** Scintillation camera, PMT

Source diaphragm distance (SDD)

(Radiotherapy) This describes the distance from the source to the forward part of the secondary collimators. It can also be called the source to collimator distance (SCD).

**Related Articles:** Source axis distance (SAD), Source surface distance (SSD)

Source loading in brachytherapy

(Radiotherapy, Brachytherapy) The brachytherapy source/s must be handled and loaded into the applicators for treatment, and many methods have been used over the time. These methods have been developed primarily to reduce the dose to the personnel, but also to improve the quality of the treatment itself:

1. Manual loading
   a. Historical method, for example handling of radium sources, which were manually introduced and removed; treatment times from a number hours to several days:
      i. Speedy insertion techniques were mandatory.
      ii. The patient was a radiation source during the long treatment.

2. Manual afterloading
   a. Still used:
      i. Applicators, needles, catheters, etc. are inserted.
      ii. Correct applicator positions are verified using dummy sources.
      iii. Improves the accuracy in applicator positioning, as there is no risk of dose to staff.
      iv. Finally, the sources are inserted into the applicators manually.

3. Remote afterloading
   a. Recommended method:
      i. Applicators, needles, catheters, etc. are inserted.
      ii. Correct positions are verified using dummy sources.
      iii. The source is loaded into the applicator(s) using a remote-controlled afterloading unit. Relevant personnel operate the afterloading unit from the operator’s room close to the treatment room (compare ‘linac’ treatments).

Note! Today, many permanent prostate implants are performed with manual loading and manual afterloading techniques, using low-energy sources. During these procedures, dose to staff from the sources used is very low; if fluoroscopy is used during the implant procedure, the dose to staff comes mainly from the fluoroscopy.

**Related Articles:** Brachytherapy, Afterloading, Manual loading, Manual afterloading, Remote afterloading, Remote afterloading unit

Source localisation

(Nuclear Medicine) In nuclear medicine imaging this refers to the process by which a gamma ray entering the gamma camera is represented at the appropriate point on the image. Please see the article Source coordinates for more information.

**Related Articles:** Source coordinates, Scintillation camera

Source models

(Radiotherapy, Brachytherapy) A source model is used in a treatment planning system for dose distribution calculations. The source model describes the dose distribution around a source and is specific
Source–skin distance (SSD)  

Source strength (brachytherapy)  

Source–skin distance (SSD)  
(Radiotherapy) See Source surface distance

Source strength (brachytherapy)  
(Radiotherapy, Brachytherapy) Calibration of source strength is a very important part of a comprehensive brachytherapy quality system. The instruments, ion-chambers and electrometers, used for source strength determinations, should have calibrations that are traceable to national and international standards.

**Specification of Source Strength for Photon Emitting Sources:** Source strength for a photon-emitting source can be given as a quantity describing the radioactivity contained in the source or as a quantity describing the output of the source:

1. Specification of contained activity
   a. Mass of radium; mg Ra
   b. Contained activity; Ci, Bq
2. Specification of output
   a. Equivalent mass of radium; mg Ra eq
   b. Apparent activity
   c. Reference exposure rate
   d. Reference air kerma rate
   e. Air kerma strength

When brachytherapy was introduced as a treatment modality, the only sources available were radium sources. Source strength was given as the mass of the radium contained in the encapsulated source. The filtration of the encapsulation was also given; usually 0.5 mm Pt for needles and 1–2 mm Pt for tubes. (The UK’s National Physical Laboratory [NPL], a standards/measurement institute established at the turn of the last century, acquired its first radium standard in 1913, made by Marie Curie, and specified in terms of mass of radium.)

Contained activity is a quantity that can be used for all types of brachytherapy sources. But, for brachytherapy dosimetry, the quantity of interest is the output of the encapsulated source, not the contained activity. (Sources are encapsulated, and it is thus difficult to determine the contained activity.) The quantity apparent activity, which is an output specification, has been used as an alternative to contained activity and is still used, especially for radiation protection applications.

The apparent activity of an encapsulated photon emitting source is the activity of a hypothetical unfiltered point source of the same nuclide that gives the same air kerma rate or exposure rate at the same distance from the centre of the source.

When artificial radionuclides became available, the brachytherapy community aimed at specifying source strength for these ‘radium substitutes’ in a radium-like manner. The new sources were similar in shape and strength to the old ones, and thus all the experience gained from the earlier radium treatments could easily be transferred.

The equivalent mass of radium, the mgRaEq, for an encapsulated photon emitting source is the mass of Ra-226 filtered by 0.5 mm Pt that gives the same air kerma rate or exposure rate at the same distance from the centre of the source. (Note that this could lead to interpretation problems for Ra sources! Consider a Ra tube with strength 20mg and filtered by 1 mm Pt, not 0.5 mm Pt; the strength of the tube will correspond to 18.7 mgRaEq!)

The ICRU Report 38 – Dose and Volume Specification for Reporting Intracavitary Therapy in Gynecology – contains the following statement:

> It is recommended that radioactive sources be specified in terms of ‘reference air kerma rate’. The reference air kerma rate of a source is the kerma rate to air, in air, at a reference distance of 1 meter, corrected for air attenuation and scattering. For this purpose, the quantity is expressed in mGy·h⁻¹ at one metre.

Task Group 43 of the AAPM defines ‘air-kerma strength’:

‘Air-kerma strength’ has units of μGy·m²·h⁻¹ and is numerically identical to the quantity Reference Air Kerma Rate recommended by ICRU 38 and ICRU 60 (ICRU 1985, 1998). For convenience these unit combinations are denoted by the symbol U where \( U = 1 \mu\text{Gy} \cdot \text{m}^2 \cdot \text{h}^-1 = 1 \text{cGy} \cdot \text{m}^2 \cdot \text{h}^-1 \). The National Institute of Science and Technology (NIST) maintains the US primary air-kerma standards for x-rays in the energy range of 10–300 keV and for photon-emitting radionuclides such as \(^{137}\text{Cs}, ^{90}\text{Sr}, ^{103}\text{Pd}\) and \(^{125}\text{I}\). Air-kerma strength, \( S_k \), is the air-kerma rate, \( K_d(d) \), in vacuo and due to photons of energy greater than, \( \delta \) at a distance \( d \), multiplied by the square of this distance, \( d^2 \):

\[
S_k = K_d(d) \cdot d^2
\]

In modern brachytherapy dosimetry, reference air kerma rate or air kerma strength is the quantity used to calculate absorbed dose. Apparent activity is still used in some treatment planning systems to specify source strength. The user of such a system is cautioned to use the same value of the air kerma rate constant as the value used in the treatment planning system, when calculating apparent activity from the measured source output, the reference air kerma rate/air kerma strength.

**Abbreviations:** ICRU = International Commission on Radiation Units and Measurements and AAPM = American Association of Physicists in Medicine.

**Related Articles:** Mass of radium, Contained activity, Equivalent mass of radium, Apparent activity, Reference air kerma rate, Air kerma strength


**Source–surface distance (SSD)**

*(Radiotherapy)* This describes the distance from the target source to the surface of the phantom (or patient). It is widely used in radiotherapy for photon beam calculations and patient set-up. It is historically used for when treatments are delivered at a fixed source to surface distance SSD, often as the distance was fixed by the use of an applicator. Standard SSD calibration uses a reference depth located at 10 cm depth in a phantom with a standard SSD equal to the source axis distance (SAD). The distance from the source to the reference point is thus equal to SAD + 10 cm. The term also used widely now in for isocentric treatments to aid patient set up.

This term can be described as source skin distance (SSD). It can be measured using the optical distance indicator (ODI) on linear accelerators.

**Abbreviations:** SSD = Source surface distance and SSD = Source skin distance.

**Related Articles:** Optical distance indicator, SAD source axis distance

**Source-to-image distance (SID)**

*(Radiotherapy)* This is used in portal imaging, and describes the distance from the source to the plane of the image. This is equivalent to source-to-film distance before the advent of portal imaging.

**Related Articles:** Source axis distance (SAD), Source surface distance (SSD)

**Space-charge effect**

*(Diagnostic Radiology)* The electrons emitted from the heated cathode filament form an electron cloud around it. This cloud is called space-charge. When the electrons leave the filament the cathode loses part of its negative charge and becomes more ‘positive’. This attracts back some of the electrons. Normally an equilibrium state exists between the number of electrons emitted and these attracted back. The electron cloud remains near the filament until high voltage (from 20 to 150 kV) is applied between the cathode and the anode. This effect is the main reason for saturation of the anode current at lower kV (usually below 50 kV), known as ‘space-charge limited’ operation of the x-ray tube *(see article on Filament current)*. The space-charge effect *(Figure S.65)* can be regulated by the Wehnelt electrode in grid-controlled x-ray tube, thus allowing the creation of very short x-ray pulses.

**Related Articles:** Cathode, Filament circuit, Filament heating, Filament current, Tube current, Wehnelt electrode

**Spatial average intensity \( I_{SA} \)**

*(Ultrasound)* The spatial average intensity is the intensity averaged over a cross-sectional area within a transmitted ultrasound field that may contain variations in intensity from point to point within the measured cross section. For example, the average intensity across a whole beam width may be measured, including the central peak intensity and the low-intensity side lobes.

The area of averaging is either (a) determined by the area of the measuring device, for example a wide hydrophone, or (b) from an integration of measurements made over a defined cross-sectional area of the transmitted field.

**Related Articles:** Time average intensity, Intensity, Beam width/area, Side lobe, Hydrophone

**Spatial filtering**

*(Nuclear Medicine)* Spatial filtering is an image processing tool used to enhance the appearance of an image. In nuclear medicine, for example images have a grainy appearance because of the statistical nature of the acquisition. A type of spatial filtering called smoothing can be used to reduce this effect, although it does adversely affect the spatial resolution of the image.

Mathematically, a filter is a matrix which is applied to all pixels in the image using an operation known as convolution. It is defined as follows:

\[
g(x) = \int_{-\infty}^{\infty} I(x') h(x - x') dx'
\]

where

- \( I(x) \) is the input image
- \( g(x) \) is the result
- \( h(x) \) is the convolution kernel which represents the filter

The convolution operation is usually denoted by \( \ast \):

\[
g(x) = I(x) \ast h(x) = h(x) \ast I(x)
\]

Instead of using convolution to apply a filter to an image, it is much easier to use Fourier transforms. This is because the Fourier transform of the convolution of two functions is equal to the product of their individual Fourier transforms:

\[
FT \{ I(x) \ast h(x) \} = FT \{ I(x) \} \times FT \{ h(x) \}
\]

**Related Articles:** Filtering, Kernel
Spatial frequency
(Magnetic Resonance) Spatial frequencies are the elements of the Fourier domain of a spatial distribution of signals. These are vectors representing a plane waves in space. They carry the unit of a wave number, that is of inverse distance. In the context of MRI this Fourier domain is called k-space. High spatial frequencies correspond to fine detail in an image (like sharp edges). Thus, the maximum spatial frequency is often used as a measure of spatial resolution.

See articles on k-space, Bar phantom and Line pair

Spatial linearity
(Magnetic Resonance) See Gradient linearity

Spatial peak intensity $I_{sp}$
(Ultrasound) The point in the ultrasound field where the highest intensity is measured. It may be measured in one instant in time or averaged over time.

The time average value (spatial peak time average intensity $I_{spTA}$) is an important quantity to know because in an attenuating medium that is the point where the maximum ultrasound power is being deposited and where heating from the ultrasound will therefore be greatest. It will usually be found at transmit focal point.

Related Articles: Intensity, Heating, Time average intensity

Spatial pulse length
(Ultrasound) Spatial pulse length is the distance that a pulse occupies in space. The spatial pulse length is equal to the number of cycles in the pulse multiplied by their wavelength. It determines axial resolution, which can be said to be approximately half the spatial pulse length due to tapering of the pulse. The spatial pulse length is inversely proportional to the frequency and directly proportional to the number of cycles in the pulse. Tapering of the pulse edges makes it necessary to determine a criterion, for example a $\sim 3$ dB limit, to compare pulses.

Spatial resolution
(Diagnostic Radiology) Spatial resolution is the ability of an imaging system to resolve, or see, the separation between two relatively small and closely spaced objects. This resolving is reduced by blurring. Resolution test objects consisting of adjacent lines separated by spaces (line pairs – spatial frequency) are used to evaluate the effect of blurring in imaging procedures.

The best quantitative descriptor of spatial resolution is the modulation transfer function (MTF). In general terms MTF is the ratio between the output and the input signal (imagined line pairs):

$$MTF = \frac{(\text{spatial frequency of the imaged signal})}{(\text{spatial frequency of the original object signal})}.$$  

The limiting spatial resolution of one system is normally taken at $MTF = 0.1$ (10%), what is also known as cut-off frequency. This way the limiting spatial resolution of a typical x-ray radiographic film is $8$–$10$lp/mm, the same figure for a typical fluoroscopic system can be $4$–$6$lp/mm.

As spatial resolution in x-ray imaging is measured by test objects with high absorption line pairs, in some cases (as in CT scanners) spatial resolution can be also named high-contrast resolution (as opposed to low-contrast resolution, what is just contrast resolution).

Related Articles: Modulation transfer function, Line pairs, Unsharpness, Contrast resolution, Bar phantom

Spatial resolution
(Nuclear Medicine) Spatial resolution refers to the detector systems’ ability to provide sharpness and details in an image.

The spatial resolution in most nuclear medicine images is somewhat limited; at least compared to the resolution in radiographic images. The spatial resolution depends on a number of parameters which differ between SPECT and PET.

Single Photon Imaging: The gamma camera spatial resolution is mainly limited by the collimator resolution and intrinsic resolution. The collimator resolution is dependent on the width and length of the collimator holes and the intrinsic resolution is affected by crystal properties and the electronics. The spatial resolution of an object is also dependant on its distance from the collimator face.

PET: In PET there are three different factors that contribute to the spatial resolution degeneration. The positron ($\beta^+$-particle) is emitted and it does not annihilate immediately. Since the positron path before the annihilation is not known, the spatial resolution can never be less than the positron range.

An assumption in PET is that when the positron annihilates the positron has no momentum, hence the annihilation photons are emitted with a $180^\circ$ opposite angle. This is untrue because in general the positron is only slowed down to thermal speeds before annihilation. As a result the positron annihilation angle deviates (~0.5°) from 180°.

Due to the depth of interaction effect the spatial resolution in PET is degraded when the source is moved further away from the centre of rotation.

To read more about the spatial resolution in PET and scintillation camera please read Related Articles.

Abbreviation: SPECT = Single photon emission computed tomography and PET = Positron emission tomography

Related Articles: Annihilation, Annihilation coincidence detection, Beta decay, Compton effect, PET, Spatial resolution PET, Spatial resolution SPECT, SPECT

Spatial resolution in a scintillation camera

Spatial resolution in a scintillation camera (Nuclear Medicine) Spatial resolution in single photon emission computed tomography (SPECT) is determined by number of factors. The two most important are the intrinsic resolution $R_{\text{int}}$, and the collimator resolution $R_{\text{coll}}$. These two factors affect the overall resolution which is called the system resolution $R_{\text{sys}}$. The system resolution, in FWHM is given by

$$R_{\text{sys}} = \sqrt{R_{\text{int}}^2 + R_{\text{coll}}^2}$$  \quad (S.19)

The collimator resolution is determined by the width and length of the collimator holes. For example, the lowest principal spatial resolution can never be lower than the hole width. The hole cannot be made too small if one expects to attain reasonable collimator efficiency. Collimators with high resolution have many small holes with thinner septa (or longer holes), but a low efficiency. Since $R_{\text{coll}}$

FIGURE S.66  Spatial resolution in the three planes (L) of an ultrasound image (R). The axial and lateral resolution are evident from the image; the effect of changes in slice thickness, which is usually the worst plane for resolution is not apparent.

FIGURE S.67  Image of an ultrasound phantom.

FIGURE S.68  The effects of improved spatial resolution is displayed in these two images of an older scanner (a) and a newer scanner (b) with improved axial and lateral resolution.
Spatial resolution PET

depends on the source to detector distance (Figure S.69) so does the system resolution.

Spatial resolution is dependent on the distance between the source and the detector. The collimator will always allow photons with an angle of incidence lower than a certain angle threshold to pass. In the left example the organs will overlap whereas in the right example when the sources are closer to the detector the images do not overlap.

When the distance between the source and the detector increases the spatial distribution of the photons is also increased. This can be seen in Figure S.69. Photons with an angle of incidence lower than a threshold angle will pass through the collimator no matter what the distance between the source and the detector is. At organ depths (5–10 cm) the $R_{sys}$ is primarily determined by $R_{coll}$ since the $R_{sys}$ is much poorer than $R_{Int}$.

Another process that has a degenerative effect on the spatial resolution is scattered photons or photons that pass right through the collimator. These photons, if registered, are interpreted as true events when in fact they are false. False events will cause blurriness and a lower SNR.

Other factors that have a degenerative effect on spatial resolution are patient movement during image acquisition and image smoothing in the preprocessing stage.

Related Articles: Annihilation, Annihilation coincidence detection, Beta decay, Compton effect, Depth-of-interaction, Intrinsic resolution, PET, SPECT


Spatial resolution PET

(Nuclear Medicine) PET spatial resolution refers to the PET detector system's ability to provide sharpness and details in an image.

The parameters affecting the spatial distribution in PET differ from SPECT. For example, the collimator resolution is not a factor in PET imaging since the annihilation coincidence detection (ACD) is used for spatial localisation. The PET system resolution is dependent upon

1. Positron physics
2. Detector properties
3. Source location relative centre of rotation

The limiting physical factor for the spatial resolution in PET is the distance travelled by the positron before annihilation. One can never attain images with lower resolution since one does not know the actual path traveled by the positron before annihilation. The maximum energy of the beta particle emitted from $^{18}$F is 635 keV which corresponds to a mean range $R_{range}$, in water of approximately 0.2 mm.

In addition to the degenerative effects of positron range, the positron is seldom brought to a full stop before annihilation, leading to a small deviation ($\pm 0.23^\circ$) from an 180° opposite emission angle. The ACD will incorrectly assign such a coincidence as a straight line of response as displayed in Figure S.71. The effect on spatial resolution, expressed in terms of FWHM, is dependent on the distance between the detectors, D and is given by

$$R_{sys} = 0.0022 \times D$$

(S.16)

For a PET system with discrete detector elements the effect on the spatial resolution, $R_{det}$, is primarily determined by the width of the detector element, $d$. $R_{det}$ is equal to the full width at half maximum (FWHM) for a point source on a line be two opposite detector element. When the point source approaches a detector element FWHM increases hence the best resolution for a PET system is in the centre of rotation (COR). In COR $R_{det}$ equals half the detector element width.

The system resolution for a source at COR ($R_{det} = d/2$) is given by

$$R_{sys} = \sqrt{R_{range}^2 + R_{coll}^2 + R_{det}^2}$$

(S.17)

If the point of annihilation is not in the centre of rotation the two photons might enter the crystal at an oblique angle. In such a case the photons can penetrate one or more adjacent crystals and interact in another crystal. Again the line of response will be misplaced and

FIGURE S.69 Scintillation camera spatial resolution at two different distances between source and detector.

FIGURE S.70 Example of speckle as seen in an ultrasound image.
as a result the spatial resolution will decrease. This is called the depth-of-interaction (DOI) effect. For a source located off-centre the DOI degeneration of spatial resolution is given by

\[
R_{\text{doi}} = R_{\text{det}} \times \left[ \cos \theta + \left( \frac{x}{d} \right) \sin \theta \right]
\]  

(S.18)

where

- \( d \) and \( x \) is the width and the length of the detector.
- \( \theta \) is the angle indicated in Figure S.72.

Another degenerative effect is false coincidences. Two examples are shown in Figure S.73. Random and scattered coincidences will yield false positional information. These random coincidences will contribute to the overall background thus leading to a loss of contrast.

While reconstructing image data it is common to use some kind of smoothing filter. This procedure can also reduce the spatial resolution.

Patient’s movement can obviously lead to a blurring of otherwise sharp edges. Therefore it is important to minimise patient movement during image acquisition.

**Abbreviation:** PET = Positron emission tomography.

**Related Articles:** Annihilation, Annihilation coincidence detection, Beta decay, Depth-of-interaction, PET, SPECT


**Spatio-temporal contrast sensitivity**

(General) Spatio-temporal contrast sensitivity refers to the perception of images which change intensity across space and over time. The effects can be demonstrated by observing a flickering grating which shows that the contrast required to detect the grating is dependent on the relationship between its spatial and temporal frequency.

**Related Article:** Perception

**Further Reading:** http://www.psypress.com/mather/resources/

**Specific absorption rate (SAR)**

(Magnetic Resonance) The specific absorption rate is defined as the radio frequency (RF) power absorbed per unit of mass of an object, and is measured in watts per kilogram (W/kg). The SAR describes the potential for heating of the patient’s tissue due to the application of the RF fields necessary for the MR measurement. The interactions of RF fields with the biological tissues and bodies depend on many parameters in a very complex way. The radio waves in free space are characterised by the intensity of the electric (E) and magnetic (H) fields and by their frequency, direction and polarisation. Inside the body the interaction with the internal RF field depends on the parameters of the external RF field as well as the electromagnetic properties of the exposed body. Since a detailed analysis of this interaction is very complex, usually simple interaction models were analysed. SAR is defined as the time derivative of the incremental energy (dW) absorbed by or dissipated in an incremental mass (dm) contained in a volume element (dV) of a given density (\( \rho \)):

\[
SAR = \frac{d}{dt} \left( \frac{dW}{dm} \right) = d \left( \frac{dW}{\rho dV} \right) \quad (W/kg)
\]

The equation indicates the rate at which the RF energy is converted into heat and provides a quantitative measure of all the interaction that depends on the intensity of the internal electric field. In MR imaging RF absorption cause a tissue heating. The patient’s ability to dissipate excess heat is an important safety issue. The temperature rise in patient resulting from a given SAR depends mostly on the blood flow and the volume of tissue receiving radiation energy and to a lesser extent on thermal conduction. SAR also depends on the pulse duty cycle, tissue density conductivity, patient size, the type and shape of the coil. Therefore the patient weight and the selected pulse sequence parameters are an important factor to be taken into account to ensure that the SAR does not exceed the permitted levels. For safety reasons, the RF power emitted by the system into the body is monitored and the respective SAR values are limited accordingly. The IEC limit values are 4 W/kg (whole body) as well as 8 W/kg (spatial peak). The FDA has established different SAR limits: 4 W/kg averaged over the whole body for any 15-min period; 3 W/kg averaged over the head for any 10-min period or 8 W/kg in any gram of tissue in the extremities for any period of 5 min. The guidelines limits for the RF patient exposure are only based upon the assumption of heating effects. If the receiving RF coil is in resonance with the transmitter, it may act to increase the RF field close to the coil. This increase in field strength is of particular concern when it occurs near the eyes. To eliminate this effect, the receive coil is decoupled during transmission. The SAR is highest with pulse
sequences that require large flip angles such as 180° RF pulses in fast spin echo sequences. MR systems calculate the SAR as its measurements is not trivial. The standard IEC 60601-2-33 establishes three modes of operation: level 0 (normal operating mode) with a SAR less than or equal to 1.5 W/kg; level I (first level-controlled operating mode) with a SAR greater than 1.5 W/kg but less than 4 W/kg and level II (second level-controlled operating mode) with the SAR greater than 4 W/kg. The limits are applied to normal environment conditions in temperature and humidity and for patient lightly clothed and resting.

Specific activity

(Nuclear Medicine) Specific activity is the amount of activity per gram of the radioisotope of interest (Bq/g) in a sample or a radioactive compound. A radioactive sample seldom or never consists of single particular isotope. Instead the sample contains a mixture of radioactive and stable isotopes. If a one or a number of stable isotopes are present in the sample, they are called carriers. A sample containing carriers is said to be with carriers. If the sample does not contain any stable isotopes the sample is carrier free. Whether or not the sample is carrier free depends on the production method. An example of a radioisotope that is never carrier free is 99mTc because the decay product is the long-lived 99Tc.

If a large quantity of an element is injected, there might be a pharmacologic response to the specific isotope. For example, too much iodine can cause a reaction. But a sample of 0.4 MBq of carrier-free 131I only contains 10−10 g of element iodine, which is not by far enough to cause an iodine reaction.

Radioactive isotopes are often attached to biological compounds in order to trace (study) a certain biological process. These compounds could be protein, glucose or antibodies. In such situations the specific activity is given by the activity per labelled substance mass (Bq/g).


Speckle

(Ultrasound)

Background: Ultrasound images of tissues show a granular appearance, which is called ‘speckle’ after a similar effect in laser optics. The granular appearance can be interpreted as noise, but is deterministic, that is if the transducer is returned to the exact same location, the exact same image will be produced. The effect can be explained as an interference phenomenon: tissue can be modelled as a collection of scatterers which are so small that a large number of them occupies one resolution cell, and all scattered wavelets from this resolution cell will interfere constructively or destructively. As high amplitude of the returning echoes is represented by white in the ultrasound image (and low by black), the result is a granular texture of the image, which notably has no correlation to the microstructure of tissue – see Figure S.70.

Speckle Statistics, First Order: The speckle noise obeys a Rayleigh probability density function if the phases are uniformly distributed between 0 and 2π and the number of scatterers is large within a resolution cell. Under those conditions, the ratio between the mean amplitude level and the standard deviation of the amplitude (commonly referred to as signal-to-noise ratio [SNR]) will be 1.91 for the received signal. Note that this is valid only for data that has not undergone any form of logarithmic or other form of nonlinear processing.

Speckle Statistics, Second Order: The size of the speckle granules is in the same order as the resolution of the imaging system, both in the lateral and the axial direction, determined by beam width and pulse length, respectively. The mathematical procedure to describe this is in terms of a correlation function, and the more mathematically stringent way to describe this is that the effective width of the correlation function corresponds to the axial and lateral resolutions. The derivation of these correlation functions is rather laborious, but is basically the result of a convolution of the point spread function and the distribution of point scatterers, followed by translations of the point spread function as the dependent variable.

Speckle Reduction: In order to reduce the speckle noise, statistically uncorrelated images of the same area can be incoherently added (compounding). Uncorrelated images can be obtained either by changing the centre frequency of the transmitted pulse, or by a given translation of the ultrasound beam. Both approaches have the effect of changing the interference pattern from the scatterers, while specular echoes will remain unaffected. It can be shown that if N images, that are independent in terms of the random speckle process, are averaged, the SNR will increase by a factor of square root of N. To obtain practically independent images the displacement of two apertures needs to be 0.4–0.5 aperture widths. For a displacement of 0.5, the amplitude correlation coefficient will be 0.25. Decorrelation may be obtained for even smaller displacements if the area is far from the focal region.

The first approach is used in beam formers, which sometimes are designed to form at least two beams of different sub-bands within the usable bandwidth of the transducer, which are then subsequently summed.

At least one manufacturer is known to use speckle reduction by phasing beams in different directions from a linear array, thus imaging the same area from different angles, and thereby different effective apertures. However, the number of beams used seems to exceed the necessary number for independent images, thus resulting in ‘redundant’ beams. This may however be necessary due to the slight distortions in tissue: if there are regions with different sound speed, sound waves are refracted. This degrades the quality of the compounded image due to angular distortion and a shift of sampling points in the individual scans. Ideally, some sort of geometric correction of individual images would thus be desirable, but the compounding seems to work well even without such correction.

Speckle decorrelation

(Ultrasound) Speckle is the granular appearance in an ultrasound image and is the result of constructive or destructive interference of echoes emanating from the same resolution cell. As long as the positions of all scattering objects are the same, the same ultrasound image will result. As the scatterers start to move relative to one another, the speckle pattern changes its appearance, or decorrelates. Conversely, the speckle pattern will also decorrelate if the scatterers are stationary, but the transducer is translated. The rate of decorrelation of the ultrasound amplitude at a certain location for a given translation is described by a correlation function in the axial, lateral, and elevational directions, respectively. The derivation of these correlation functions is rather laborious, but is basically the result of a convolution of the point spread function and the distribution of point scatterers, followed by translations of the point spread function as the dependent variable. To obtain practically independent amplitudes at a certain location, the displacement of two apertures needs to be 0.4–0.5 aperture widths. For a displacement of 0.5, the amplitude correlation coefficient will be 0.25. Decorrelation may be obtained for even smaller displacements if the area is far from the focal region.

SPECT (single-photon emission computed tomography)

(Nuclear Medicine) See Single photon emission computed tomography (SPECT)
TABLE S.2
Common Radionuclides in SPECT Imaging and Their Applications

<table>
<thead>
<tr>
<th>Radionuclide</th>
<th>Half-Life</th>
<th>Molecular Imaging Applications</th>
</tr>
</thead>
<tbody>
<tr>
<td>⁹⁹mTc</td>
<td>6h</td>
<td>Tumour detection and characterisation, cardiac infarction detection and</td>
</tr>
<tr>
<td></td>
<td></td>
<td>monitoring thrombolytic therapy, renal function studies</td>
</tr>
<tr>
<td>¹³¹I</td>
<td>8 days</td>
<td>Thyroid function and tumour detection, renal function studies, receptor binding</td>
</tr>
<tr>
<td></td>
<td></td>
<td>studies, transporter function</td>
</tr>
<tr>
<td>¹¹۱In</td>
<td>2.8 days</td>
<td>Inflammatory disease detection, neuroendocrine tumour detection, receptor</td>
</tr>
<tr>
<td></td>
<td></td>
<td>binding studies</td>
</tr>
</tbody>
</table>

SPECT clinical applications

(Nuclear Medicine) The clinical uses of single photon emission computed tomography (SPECT). SPECT is the most widely used nuclear medicine technique for clinical imaging. The radionuclides used for SPECT images typically have half-lives ranging from 6h to 8 days. Such radionuclides can be manufactured at a remote facility and then transported to the users location, thus there is no need for expensive radionuclide production equipment, for example cyclotrons. Below is a table with radionuclides used for SPECT imaging and their clinical application (Table S.2).

Abbreviation: SPECT = Single photon emission computed tomography.


SPECT-CT scanner

(Nuclear Medicine) A combination of a single photon emission computed tomography (SPECT) scanner and a computed tomography (CT) scanner. The SPECT scanner acquires images of the in vivo radiotracer distribution while the CT scanner provides morphological images. The combination of these two scanners allows accurate localisation of abnormalities imaged by the SPECT.

The CT data can also be used to correct for photon attenuation in the patient body.


Spectra

(Magnetic Resonance) The term spectra (singular spectrum) refers to signals that have been decomposed into components with different frequencies, often specifically to a two-dimensional plot of signal intensity against frequency.

In the present context, an NMR spectrum is obtained by Fourier transformation of the time-domain NMR signal (Figure S.74).

Related Article: Magnetic resonance spectroscopy, MRS

Spectral analysis

(Magnetic Resonance) This term refers to the steps that are applied to analyse in vivo NMR spectra in order to yield qualitative or quantitative results.

In vivo spectra present a number of challenges in this regard, as compared to in vivo data. Peaks are generally broad and frequently overlap: it may be impossible to resolve closely spaced peaks due to different compounds, and multiplet structure that is well resolved in vivo is often lost. In phosphorus (³¹P) spectra in particular, there are often broad baseline features due to membrane phospholipids.

This article presents an overview of the main stages of spectral analysis. Certain key stages are described in more detail in dedicated articles.

Time Domain Processing: These steps are applied to raw data in the time domain, prior to Fourier transformation.

- DC Correction: The mean value of a section of the spectrum well away from any resonance peaks is subtracted from all of the data points. This eliminates any offset voltage that would otherwise appear as a zero-frequency spike in the frequency domain spectrum.
- Sensitivity Enhancement/Line Broadening: The time domain signal is multiplied by a decaying function, frequently an exponential, in order to weight out later data points that contribute to noise but not to signal. An undesirable, but inevitable, consequence is that lines in the spectrum are broadened, so there is a trade-off between improvement in signal-to-noise and loss of resolution.
- Resolution Enhancement: This can be achieved either by multiplying the time domain signal by a rising function (e.g. a negative exponential), at the cost of reduced signal-to-noise, or by zero filling the data up to the next power of two. This can allow previously unresolved multiplet structure to be recovered.
- Baseline Correction: Some baseline correction methods operate on time domain data. Please refer to article Baseline correction for details.

Frequency Domain Processing: These steps are applied in the frequency domain, after Fourier transformation.

- Baseline Correction: Some baseline correction methods operate on frequency domain data. Please refer to article Baseline correction for details.
- Phasing: The delay between nuclear excitation and signal acquisition allows the development of both frequency-independent and frequency-dependent phase. These effects can be eliminated by manual adjustment of phase, achieved while viewing the spectrum at the spectrometer console,
or increasingly by less onerous, automated means. The latter can include corrections derived from the unsuppressed water peak, or calculated by numerical optimisation.

**Quantification:** Measurement of the areas of the resonance peaks in a spectrum may be achieved using a number of time-domain and frequency-domain techniques. See article Peak areas for details.

Peak areas may be converted into absolute concentrations by comparison with an "internal standard" (a compound within the tissue of known or assumed concentration), by using an external standard placed adjacent to the subject's body or studied in a separate experiment, or by using an external standard to determine the concentration of an internal marker (often water) which can then itself be used as an internal standard. If separate experiments are performed, great care is needed to ensure that parameters such as receiver gain are properly accounted for. Relaxation correction, using either literature values or values measured in situ, is a requirement of absolute quantification.

**Related Articles:** Baseline correction, Peak areas, Peak assignment

**Spectral broadening** *(Ultrasound)* The frequency shift due to the Doppler effect is given by the Doppler equation. Clearly, if all targets (red blood cells) within the ultrasound beam have the same velocity (a situation that occurs in plug flow) and if the angle is fixed (a single value for $\theta$) then the spectrum of the Doppler signal will be a single spectral line. Spectral broadening will occur if either the beam is wide, so that the angle subtended by the beam is not a single value (the so-called geometrical spectral broadening), or if the distribution of velocities of the red blood cells within the beam is wide (i.e. there is a statistical blood velocity distribution different from that of plug flow). The beam can be made narrow, so the geometrical factor can be controlled. If that is done, then spectral broadening is an indication of a broad profile of blood velocities.

Spectral broadening is associated with turbulence in arterial sites just distal to plaques, for instance, and this is used in clinical practice: At the site of the plaque, the stenosis results in an increased blood velocity and at sites just distal from the plaque, spectral broadening occurs as a consequence to the disturbance in the flow – which might include reverse flow in certain areas.

**Spectral display** *(Ultrasound)* See Sonogram

**Spectral matching** *(Magnetic Resonance)* This term refers to the identification of resonance peaks in an NMR spectrum. It is a synonym for peak assignment.

**Related Article:** Peak assignment

**Spectral width** *(Magnetic Resonance)* This term refers to the range of frequencies present in a spectrum. The spectral width required in a particular experiment is determined by the range of frequencies over which spectral peaks are expected to occur.

The NMR signal induced in the receive coil is sampled and converted by an analogue-to-digital converter. According to the Nyquist theorem, the ADC sampling rate must be at least twice the highest frequency present in the signal if aliasing is to be avoided.

In order to satisfy the Nyquist theorem, the sampling rate, $f$, must be set so that

$$f = 2\Delta\omega$$

where $\Delta\omega$ is the spectral width.

Filters are normally used to ensure that noise (and any unanticipated signals) outside the spectral width are not aliased back into the spectrum.

**Related Article:** Nyquist theorem

**Spectroscopic imaging** *(Magnetic Resonance)* See Chemical shift imaging (CSI)

**Spectrum, continuous** *(General)* A continuous spectrum is a spectrum where energy is produced at all possible wavelengths between certain specified minimum and maximum values. The most common example of a continuous spectrum (or thermal spectrum) is blackbody radiation where the spectrum from an ideal radiator or absorber depends only on temperature. The continuous spectrum is also sometimes known as a thermal spectrum as hot, dense objects will emit electromagnetic radiation at all wavelengths. Examples of a continuous spectrum are the Sun and a rainbow.

Another source of a continuous spectrum, not associated with heat, is x-ray production through the rapid deceleration of electrons (also known as bremsstrahlung or braking radiation). In an x-ray tube electrons are accelerated by a voltage and strike a metal target with the x-rays produced over a range in energy from zero up to a maximum corresponding to the tube voltage.

An illustration of a continuous x-ray spectrum is given on Figure S.75.

**Related Articles:** Spectrum, Discrete

**Spectrum, discrete** *(General)* A discrete spectrum is a spectrum where energy is only produced at certain wavelengths which are characteristic of the system. Familiar examples of discrete spectra are the different coloured advertising signs (filled with different fluorescent gases), coloured street lamps and the specific colours of lasers.

This discrete spectrum arises due to the structure of atoms where internal energy can only be changed by certain set (discrete) amounts, and these discrete energies are known as quantum states. When changing between these quantum states a certain amount of electromagnetic radiation will be emitted (known as a quantum), which is equal to the energy difference between the two states. Therefore different chemical elements within the periodic table will produce different discrete spectra. Therefore it is possible to identify the chemical composition of a material emitting or absorbing radiation by measuring the spectrum of light being emitted or absorbed.

In the production of x-rays from an x-ray tube the spectrum will contain both a continuous and a discrete component. The discrete part of the spectrum arises due to electrons having sufficient energy to remove inner shell electrons from the atom. As a result of a gap in the inner shell, electrons from the higher energy outer shells will fill the gap and as a result the residual energy will be emitted in the

---

**Figure S.75** An illustration of a continuous spectrum.
Speed of film

(Diagnostic Radiology) Speed is a term used for both photographic and radiographic film describing the amount of exposure, at least in relative values, required to produce an image. Film sensitivity is an alternative quantity for film speed and is usually expressed in actual exposure values.

Generally speaking, the thicker the emulsion of the x-ray film and the larger the AgBr crystals in it, the less radiation is necessary for producing a specific radiographic optical density on the film. Such film is referred to as more sensitive, or with higher speed (as it will produce the image for shorter time). However, due to the larger crystals, such film will produce a courser image, that is with lower spatial resolution.

Film speed is measured with ASA or DIN, or ISO. For example, a film ASA 400 is more sensitive (with higher speed, but with lower resolution) than film ASA 100, etc.

Related Articles: ASA, Film type

Speed of sound

(Ultrasound) A sound wave travels with a specific speed dependent on the density and stiffness of the medium through which it is travelling. The speed of sound, \( c \), is given by:

\[
c = \sqrt{\frac{k}{\rho}}
\]

Consider a material consisting of particles with mass \( m \) linked together with springs with stiffness \( k \), Figure S.77. If the particles to the left are displaced (by a transducer) this movement will be transferred through the material. If the springs are very stiff (high \( k \)) the movement of all masses will be almost simultaneous, thus generating very high speed of sound. If the springs are weak (low \( k \)) the movement of the adjacent mass will be time shifted thus generating a lower speed of sound. A similar observation with large and small masses shows that it is easier to move small masses than large masses and that small masses will generate higher speed of sound.

\[
D_m = J_{\text{air}} \left( \frac{W}{e} \right) \left( \frac{S}{\rho} \right)_{\text{m, air}}
\]

where \( J_{\text{air}} \) is the ionisation charge per unit mass of air in the cavity \( (S/\rho)_{\text{m, air}} \) is the ratio of the mean collision mass stopping power of material \( m \) to that of air
\( W/e \) is the quotient of the average energy expended to produce an ion pair by the electronic charge

To estimate the mass stopping power ratio required in the equation, it is necessary to calculate the mean collision stopping powers for material \( m \) and air.

### Related Articles:
- Acoustic impedance, Reflection
- Spencer Attix theory

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**FIGURE S.76** An illustration of an x-ray spectrum containing both continuous and discrete components.

**FIGURE S.77** A material can be described as masses, which are connected to each other with springs, the springs represent the stiffness of the material and the masses represent the density. If the material is stiff the sound wave travels rapidly from one side to the other; if the material is dense then it is more resistant to movement by the sound wave.
the appropriate materials averaged over the spectrum of all the electrons crossing the cavity taking into account the polarisation effect and considering the effect of large discrete energy transfer with the production of $\delta$-rays. Spencer and Attix theory made allowance for $\delta$-rays by including into the so-called restricted mass collision stopping powers only energy transfers below some selected maximum $\Delta$ in the energy transfer. Electron collisions in which the energy transfer exceeds $\Delta$ are not considered dissipative and these electrons are added to the electron energy spectrum. The difference between unrestricted and restricted collision mass energy stopping powers is small, less than 1%, for all materials of low atomic number for electron energies below 1 MeV but is significant at higher energies.

Consequently the Spencer Attix cavity theory relates the dose delivered to the gas in the ionisation chamber $D_{\text{gas}}$ to the dose in the surrounding medium $D_{\text{med}}$ by the relationship

$$D_{\text{med}} = D_{\text{gas}} \left( \frac{\bar{L}}{p} \right)_{\text{med}}$$

where the mass stopping power ratio $\left( \frac{\bar{L}}{p} \right)_{\text{med}}$ is the ratio of the spectrum averaged mass collision stopping powers for the medium to that for the gas where the averaging extends from a minimum energy $\Delta$ to the maximum electron energy in the spectrum. The fundamental assumptions of the theory are that the cavity does not change the electron energy spectrum in the medium, all the dose in the cavity comes from electrons entering the cavity and that electrons with energy below $\Delta$ are in charged particle equilibrium. The theory applies where charged particle equilibrium of the electrons above $\Delta$ does not exist which is generally the case near an interface between media or at the edge of a beam.

**Related Articles:** Bragg–Gray cavity theory, Charge particle equilibrium

**SPGR (spoiled gradient recalled acquisition in the steady state)**
*(Magnetic Resonance)* See Spoiled gradient recalled acquisition in the steady state (SPGR)

**Spikes**
*(Magnetic Resonance)* Magnetic resonance images are reconstructed from data in $k$-space. If one or more of the data points in $k$-space becomes corrupt then a spike occurs. The corrupt points in $k$-space are referred to as spikes due to the fact that corrupted data points have a high signal magnitude. These spikes in the data lead to an image artefact called a herring bone artefact, where, when Fourier transformed, the spikes are convolved with the image information providing a regular series of high- and low-intensity stripes in one or more directions across the image.

There are many possible sources that will lead to the corruption of the data. These are static discharge caused by synthetic fibres near the receiver coil, mechanical stress on gradient coils, hardware, such as the RF amplifier and the gradient amplifier, failing and leaks in the shielding around the magnet.

**Abbreviation:** RF = Radiofrequency.

**Related Article:** Herring bone artefact

**Spin**
*(Magnetic Resonance)* Spin is a purely quantum mechanical form of angular momentum, arising out of the Dirac equation for relativistic quantum mechanics. The closest analogy in classical physics is the ‘spin’ of a top.

Each elementary particle, such as the electron or photon, and every atomic nucleus can be associated with it a spin quantum number $I$, such that the magnitude of its spin angular momentum is given by

$$|\vec{p}| = h\sqrt{I(I + 1)}$$

The value of $I$ is an intrinsic property of the elementary particle, similar to its electric charge or rest mass. Composite particles, such as the proton, are also described as having a fixed value of $I$, where this refers to the spin of the lowest energy state of the composite particle.

Unlike the orbital angular momentum quantum number, which may only take integer values, $I$ may also take half integer values. This leads to the classification of two types of particle, those with integer $I$ values, called bosons, and those with odd half-integer values, called fermions. An example of a boson is the photon, with $I = 1$, while both the electron and the proton are fermions with $I = 1/2$.

A particle with spin angular momentum $\vec{P}$ also has a corresponding intrinsic magnetic moment given by

$$\vec{\mu} = \gamma \vec{P}$$

where $\gamma$ is a particle specific quantity called the gyromagnetic ratio. Thus, only nuclei with non-zero spin exhibit magnetic resonance.

For a particle placed in a magnetic field, this magnetic moment will lead to an interaction energy ($E = -\vec{\mu} \cdot \vec{B}$) dependent on the projection of the spin angular momentum vector onto the magnetic field direction. The magnetic field is usually considered to define the $z$-axis, so this projection is labelled by the $P_z$ quantum number. The possible values of $P_z$ are given by

$$P_z = \hbar m_I$$

where

$$m_I = I, (I - 1), (I - 2), \ldots, -I$$

For spin $1/2$ particles like the proton, there are two possible values for $P_z = \pm \hbar/2$. This means that there are two possible stationary states, one with a lower energy of interaction than the other (see Figure S.78). This leads to a preference for the spin to be aligned in the lower energy configuration, although thermal effects mean that at room temperature the difference in number between spins at the two different energies is only about 1/1000 spins.

**Spin density**
*(Magnetic Resonance)* In MRI, spin density refers to the number of hydrogen nuclei per unit volume, precessing at the Larmor frequency. The spin density ($\rho_s$) is proportional to the magnetisation along the external magnetic field ($M_z$).

**Related Articles:** Proton density

![Figure S.78](image)

**FIGURE S.78** The energy of a spin-1/2 magnetic moment $\mu$ split by a $B$-field as seen in quantum mechanics.
Spin echo

(Magnetic Resonance) One of the most robust and frequently used pulse sequences in MRI is the spin echo (SE) sequence, introduced by Hahn long before the advent of MRI itself (Hahn 1950). In this sequence, two RF pulses are used (Figure S.79). Shaded areas in the slice and read gradient directions are refocusing pulses (see Pulse sequence). In the phase direction, different values of the gradient are used after each repetition interval in order to cover k-space.

The first RF pulse (90°) is used to flip the longitudinal magnetisation ($M_z$) into the transverse plane. The amplitude of the resulting transversal magnetisation $M_y$ is rapidly reduced by spin dephasing due to the combined effects of $T_1$ decay and influence of magnetic field inhomogeneities. Such inhomogeneities can, for example be caused by field variations in interfaces between tissues with different magnetic field susceptibility, by metallic implants and by imperfections in the main magnetic field. In order to reduce effects of field inhomogeneity, a second RF pulse (180°) is applied at time interval ($TE/2$) to reverse the order of the dephased spins. At a time interval TE the spins are refocused with respect to the field inhomogeneity effects and thus the amplitude of $M_y$ will depend only upon $T_1$ relaxation. After a given repetition time TR, when $M_y$ is recovering due to $T_1$ relaxation, the sequence is repeated with a new value of the phase encoding gradient amplitude to obtain data of the complete k-space.

The pulse created by the repetition time is normally used to excite slices at other positions, in order to facilitate multislice imaging. When using the SE sequence, the signal amplitude depends upon three object parameters (proton density (PD), $T_1$ and $T_2$) and two acquisition parameters (TE and TR) and if TR ≪ TR the signal can be approximated by Equation S.20:

$$S_{SE} \sim PD \cdot (1 - e^{-TR/T_1}) \cdot e^{-TE/T_2} \quad \text{(S.20)}$$

In standard MRI, the acquisition parameters are used to manipulate the image contrast. For example, in order to reduce the influence of $T_1$ on the signal in Equation S.20, a low value of TE relative to $T_2$, giving a low TE/$T_2$ ratio, is chosen. The following ‘rules of thumb’ are frequently used:

<table>
<thead>
<tr>
<th>PDW (PD-Weighted)</th>
<th>$T_1$W ($T_1$-Weighted)</th>
<th>$T_2$W ($T_2$-Weighted)</th>
</tr>
</thead>
<tbody>
<tr>
<td>TR ‘High’</td>
<td>‘Low’</td>
<td>‘High’</td>
</tr>
<tr>
<td>~2000 ms</td>
<td>~500 ms</td>
<td>~2000 ms</td>
</tr>
<tr>
<td>TE ‘Low’</td>
<td>‘Low’</td>
<td>‘High’</td>
</tr>
<tr>
<td>~20 ms</td>
<td>~20 ms</td>
<td>~100 ms</td>
</tr>
</tbody>
</table>

It must be emphasised, that the values given above are only valid for a certain interval of relaxation times in the object and tissue with extreme relaxation times obtain signal values that depend significantly upon several parameters in the equation.

The SE pulse sequence has the advantage of robustness and it gives large possibilities to vary object contrast by simple changes of system parameters. However, one of the drawbacks is long acquisition times. The total acquisition time of a SE sequence can be calculated by the formula

$$T_{acq, SE} = N_{ave} \cdot N_{phase} \cdot TR$$

where $N_{ave}$ denotes number of execution averages $N_{phase}$ denotes number of repetitions required for the different the phase encoding steps (number of k-space lines)

From this formula, it can easily be seen that a PD or T2W image with 256 phase steps and 2 averages takes over 17 min to acquire. Therefore much effort has been put down to further reduce acquisition times, for example by using fast spin echo sequences.

**Related Articles:** Pulse sequence, RF pulse, $T_1$, $T_2$-weighted, $T_2$, $T_1$-weighted, Fast spin echo


Spin temperature

(Magnetic Resonance) The spin temperature is used to characterise a spin population of the various energy states of the spin system that follows a Boltzmann distribution with this particular temperature, that is the spin temperature is the Boltzmann temperature that corresponds to the observed distribution of the spins. The spin temperature provides a convenient way to describe the orientational order in the spin system even when it is not in thermal equilibrium with the lattice.

**Related Articles:** Boltzmann distribution, Relaxation


Spin warp imaging

(Magnetic Resonance) In conventional Fourier transform imaging techniques, spatial information is encoded into the NMR signal by means of frequency encoding and phase encoding, producing a two-dimensional array of data in k-space from which the image can be recovered by Fourier transformation. Acquisition of this array involves repetition of the pulse sequence a number of times, with each repetition utilising a different displacement of the data acquisition trajectory along the phase encoding axis in k-space. The method of incremental displacement in almost universal use today is that known as **spin warp imaging**.

In the original implementation of this technique, due to Kumar et al., incrementation of data acquisition along the phase encoding axis was achieved by altering the duration of the phase encoding gradient, so that in each repetition magnetisation evolved in the presence of the gradient for a different period of time and hence acquired a different phase distribution, corresponding to a different value of $k$. The drawback of this technique was that the interval between excitation and signal detection varied between repetitions as well, leading to image degradation due to variations in $T_2$ relaxation and field inhomogeneity effects.
The solution, proposed by Edelstein et al. (1980) and known as spin warp imaging, was to increment the amplitude of the phase encoding gradient, rather than its duration. Because the phase accumulated by magnetisation at a given location along the gradient depends on the product of the gradient amplitude and duration, this has the same effects as the original technique of Kumar et al. (1975) while avoiding the disadvantages (Figure S.80).

**Related Articles:** Frequency encoding, k-space, Phase encoding


**Spine coil**

(Magnetic Resonance) A spine coil is used in MRI as the dedicated receive RF coil for spinal imaging. The spine coil fits to or forms part of the MRI patient table (Figures S.81 and S.82). The patient lies supine on the coil for imaging. A spine coil is a type of surface coil, providing good SNR for structures close to the coil but a limited field view in the anterior-posterior direction but large field of view along the feet-head direction.

**Spin-lattice relaxation**

(Magnetic Resonance) When an RF excitation pulse is switched off, nuclei start to dissipate energy to their surroundings (mainly via thermal mechanisms): this is spin-lattice relaxation.

**Related Article:** Relaxation

**Spin-spin coupling**

(Magnetic Resonance) Two nuclei with non-zero spin can interact to cause further energy-level splitting in the presence of an external magnetic field: this is spin-spin coupling.

**Related Article:** Magnetic coupling

**Spin-spin relaxation**

(Magnetic Resonance) This term refers to the stochastic dephasing phenomenon whereby transverse magnetisation undergoes exponential decay. Transverse magnetisation is composed of nuclei that have been put into phase coherence by a radiofrequency (RF) pulse.

Spin-spin relaxation occurs when a nucleus in the higher energy level releases energy, thus moving to the lower energy level, and this quantum of energy is absorbed by another nucleus which is thus promoted to the higher energy level. The two nuclei have in effect ‘swapped’ energy levels, but the newly excited nucleus is out of phase with those that were excited by the initial RF pulse, and as a consequence the net transverse magnetisation is smaller.

This energy-level transition does not in general occur spontaneously, but rather is induced by time-varying electromagnetic fields. For the hydrogen nuclei (protons) in a water molecule (almost exclusively the target of conventional MRI), these fields arise as a result of thermal motion of the molecule, mainly in the form of rotation, sweeping each hydrogen nucleus through the spatially inhomogeneous magnetic field generated by the other one. These are known as intramolecular dipole-dipole interactions, and fall into two categories in terms of their impact on nuclear relaxation.

Dynamic dipole-dipole effects relate to magnetic field variation which is at the Larmor frequency of the nucleus, and therefore induces energy-level transitions. The emitted energy may be absorbed by another nucleus or lost to the lattice, and so these effects contribute to both spin-spin and spin-lattice relaxation.

Static dipole-dipole effects relate to magnetic field variations that are slow on the NMR timescale (<100kHz) and so generate quasi-static inhomogeneities that cause irreversible dephasing of transverse magnetisation. These effects contribute to spin-spin relaxation only. Thus, large quasi-stationary molecules experience a high degree of static dephasing and have very short T₂, rendering them ‘MRI-invisible’.

The gross effect of spin-spin relaxation is exponential decay of transverse magnetisation (Mₓ) at a rate determined by the spin-spin or transverse relaxation time, T₂:

\[ M_\text{x}(t) = M_0 \exp(-t/T_2) \]

The freedom of molecules to move, and hence the proportion of molecules contributing to dynamic and static dipole-dipole interactions, depends on the physicochemical environment of the molecules. The resulting difference in T₂ relaxation times between water protons in different body tissues is one of the principal sources of contrast in MRI. Additional dephasing due to static field
inhomogeneities (see $T_2^*$) can be reversed by acquiring a spin echo. By varying the interval after the excitation pulse at which the NMR signal (spin echo) is collected, one can achieve different degrees of $T_2$ weighting in the resulting image.

**Related Articles:** Spin-lattice relaxation, $T_2$-weighted, $T_2^*$

**Spiral (helical) interpolation**

*(Diagnostic Radiology)* Spiral or helical interpolation is the process by which the data required to reconstruct an axial, planar image is obtained from a helical CT scan. When scanning in helical mode the attenuation data acquired at different angles in a rotation is non-planar (Figure S.83). It is therefore necessary to interpolate the acquired data in order to obtain a planar data set for image reconstruction.

Different types of interpolation can be used. On single slice scanners, the simplest is the so-called 360° linear interpolation (Figure S.84), where attenuation data at each angular position is obtained by interpolation of the two nearest measurements, obtained at the same angular tube position, on each side of the reconstruction plane. The measured data is weighted linearly with distance from the reconstruction point. More commonly on single slice systems, 180° linear interpolation is used. This utilizes the concept of complimentary projections, where attenuation data from opposing projections (e.g. anterior-posterior and posterior-anterior is regarded as equivalent). This reduces the interpolation distance and results in less broadening of the slice profile.

On multislice scanners different helical interpolation approaches are used in helical scanning. For each rotation a number of data points are available for interpolation at each angle (Figure S.85a). By selecting a filter width and only using the points within this width, different slice widths can be reconstructed from a given acquired data set. A wide filter width will result in a wider reconstructed slice (Figure S.85b), whereas a narrower filter width will give a narrow slice (Figure S.85c).

**Related Articles:** Helical scanning, Helical pitch, Slice profile, Multislice CT

**Spiral imaging**

*(Diagnostic Radiology)* Spiral imaging (or scanning) is also known as Helical scanning – see the article Helical scanning.

**Related Articles:** Helical pitch, Helical interpolation, Slips, Image artefact, CT reconstruction, Slice thickness

**Spiral imaging**

*(Magnetic Resonance)* In MR imaging, the acquired data is collected in the Fourier domain, or the so-called $k$-space. In order to form an image all data points in this $k$-space have to be obtained (e.g. sampled). The traditional way to collect data is to traverse $k$-space line by line on a Cartesian grid. However, it is also possible to collect data on other trajectories, for example by starting in the centre of $k$-space and collect outwards in $k$-space in a spiral manner. This is done by oscillating the gradient waveforms in $x$- and $y$-direction simultaneously (Figure S.86).

With a spiral acquisition scheme, the effective echo time becomes short since $k$-centre is acquired first. Spiral imaging can be used when fast collection of data is necessary, for example in dynamic MR sequences.

With spiral acquisitions, the data is usually resampled on a Cartesian grid and performing a fast Fourier transform subsequently. Spiral acquisitions are often sensitive to off-resonance effects, causing blurring in the resulting MR images.

**Related Article:** Spiral scanning


![Figure S.83 Helical path of x-ray beam around patient.](image1)

![Figure S.84 Obtaining an axial (planar) slice from a helical data set by interpolation. (Graphs courtesy of ImPACT, UK, www.impactscan.org)](image2)

![Figure S.85 Helical interpolation on a multislice scanner with variable filter width. (a) Multiple number of data points available for interpolation at each projection angle. (b) Wide filter width selected for wide reconstructed slices. (c) Narrow filter width selected for narrow reconstructed slices. (Graphs courtesy of ImPACT, UK, www.impactscan.org)](image3)
Spiral sampling

*Diagnostic Radiology*)

Spiral sampling, or *z*-sampling, refers to the sampling of data along the scan axis, or *z*-axis, of a CT scanner in spiral (helical) scanning. Figure S.87 is an example on a 4-slice CT scanner. The solid lines show the central ray paths from the four detector rows. The dashed lines are the complementary data sets for each row, synthesised from the opposing projections principle (see *Spiral (helical) interpolation*). By utilising these complementary data sets, the sampling interval may be reduced and image quality improved. On multislice CT scanners different *z*-sampling patterns arise depending on pitch employed. For integer pitch values the complementary data overlap with the direct data and no improvement in *z*-sampling is achieved (Figure S.87a). For non-integer pitch values the distance between the data points is decreased (Figure S.87b), resulting in reduced helical interpolation artefacts.

Another approach to increasing *z*-sampling is through the use of a *z*-flying focal spot (Figure S.88). The use of a flying focal spot in the scan plane to improve spatial resolution has been explained elsewhere (see *Flying focal spot*). In a similar way, by electromagnetically switching the position of the focal spot in the *z*-direction everytime the detectors are sampled, each detector row is double-sampled, halving the sampling distance and giving rise to twice the number of ‘slices’ as there are detector rows.

**Related Articles:** Helical pitch, Helical interpolation, Image artefact, Flying focal spot

**Spiral scanning**

(Diagnostic Radiology) Spiral CT scanning is also known as Helical CT scanning – see the article *Helical scanning*.

**Related Articles:** Helical pitch, Helical interpolation, Slip rings, Image artefact, CT reconstruction, Slice thickness

**Spiral scanning**

(Magnetic Resonance) During MR imaging, data is collected in Fourier space (often called *k*-space). The information in *k*-space is most often collected by acquiring the data in one or more rows during a repetition time (depending on pulse sequence), until all data in *k*-space is collected.

However, there is also a possibility to collect the data using a spiral trajectory through *k*-space, i.e. to collect data beginning from the centre of *k*-space and spiralling outwards instead of collecting the data row by row. In this so-called spiral scanning, the traditional phase- and frequency-encoding gradients are replaced by two oscillating read-out gradients during data collection (see Figure S.89).

Spiral scanning can be used in most of the MR applications in which fast acquisition of the data is desirable, for example in diffusion imaging, flow imaging, and imaging of dynamic processes.

Another advantage of spiral scanning is the more effective traversal of *k*-space, as the inner parts of *k*-space can be sampled denser. However, spiral scanning is sensitive to off-resonance effects, causing the resulting images to be blurred. For image reconstruction, the obtained data is usually resampled onto a Cartesian grid in order to perform a fast Fourier transform for the final image

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**FIGURE S.86** Readout scheme in *k*-space during a spiral acquisition. The spiral readout scheme collects data along a spiral trajectory.

**FIGURE S.87** Effect of pitch on spiral sampling: example for 4-slice CT scanner. (a) Pitch 1: For integer pitch values complementary projection data overlaps with actual projection data. (b) Pitch 0.875: For non-integer pitch values improved *z*-sampling can be achieved. (Graphs courtesy of ImPACT, UK, www.impactscan.org)

**FIGURE S.88** Diagram showing the principle of the *z*-flying focal spot. (Courtesy of Siemens.)
Spoiled gradient recalled acquisition in the steady state (SPGR)

(Magnetic Resonance) Spoiled gradient echo is a steady-state sequence that goes by a variety of names depending on the manufacturer, that is SPGR (General Electric), spoiled fast low-angle shot FLASH (Siemens) and RF-spoiled field echo (Toshiba). A steady state acquisition means that the longitudinal magnetisation $M_z$ is equal in value before each excitation pulse after a sufficient number of repetitions. The required number of start-up cycles to reach the steady state depends on TR, the longitudinal relaxation time and RF-spoiled field echo (Toshiba). A steady state acquisition means that the longitudinal magnetisation $M_z$ is equal in value before each excitation pulse after a sufficient number of repetitions. The required number of start-up cycles to reach the steady state depends on TR, the longitudinal relaxation time and flip angle $\theta$. The spoiling is introduced to force transverse magnetisation $M_x$, to be equal to zero prior to the applying of each new excitation pulse by applying radio frequency (RF) spoiling. That is done by linearly varying the phase of the RF pulse in each cycle depending of RF pulse number $n$. The phase $\varphi_{RF}$ of RF pulse must be of the form

$$\varphi_{RF}(n) = \varphi_{RF}(n-1) + n\varphi_{RF}(0) \quad n = 1, 2, 3, ...$$

The start phase value $\varphi(0)$ is recommended by Zur et al. (1991) to be 117°; this value makes the RF spoiled signal similar to the ideal incoherent (non-naturally) spoiled steady state, and is valid for all values of $T_1$, $T_2$, TR, and flip angle $\theta$ of practical interest.

Related Articles: Gradient spoiling, Steady-state


Spot test

(Nuclear Medicine) Spot test can be used to investigate the presence of aluminium in the $^{99m}$Tc-eluete. Aluminium cations are formed during the production of the $^{99m}$Tc-eluete and may be found in the eluate. The aluminium breakthrough may interfere with the $^{99m}$Tc-$^{99}$Mo-elutiation procedure and, for example cause agglutination of red blood cells during the $^{99m}$Tc-$^{99}$Mo-elutiation of the cells. Precipitation of phosphate buffer in colloid preparations and $^{99m}$Tc-$^{99}$Mo- MDP may form colloids with liver and spleen uptake and particles can be trapped in the lungs as a result.

The presence of aluminium can be detected using test kits consisting of a strip impregnated with a colour complexing agent and a reference aluminium solution. A small drop of the eluate is applied to the strip and the obtained colour of the strip is compared with the test reference solution. A more dense colour of the test solution indicates too high amount of aluminium.

Related Articles: TLC, Rf-value

Spurious echoes

(Magnetic Resonance) The term ‘echo’ in MRI refers to a signal that has been delayed (or recalled) in order to allow time for additional pulse sequence elements prior to data acquisition, and/or to introduce signal weighting. Echoes may be obtained by using RF pulses to flip magnetisation that is dephasing because of static field inhomogeneity to the opposite side of the transverse plane (a spin echo, usually obtained using a 90° and 180° pulse combination). Alternatively, signal may be deliberately dephased using a switched gradient, the effect of which is reversed by a gradient of opposite polarity at the desired later point in time (a gradient echo).

In addition to intentional generation of spin and gradient echoes, the extensive use that is made of RF pulses and switched gradients in MRI pulse sequences means that echoes may sometimes be generated unintentionally due to inopportune combinations of gradients or RF pulses. For example, a train of any three RF pulses with flip angles other than 180° will generate a type of spin echo known as a stimulated echo. An MRI experiment consisting of repetitive application of a sequence containing several RF pulses will therefore unintentionally generate many stimulated echoes as well as the intended signal. Similarly, the cumulative effect of a train of gradient pulses will in general lead to inadvertent rephasing of magnetisation, generating unwanted gradient echoes.

These unintentional echoes may collectively be termed spurious echoes. The phenomenon can become problematic if spurious echoes, which may not have experienced the same history of spatial encoding as the intended signal and may have different contrast properties, fall within the data acquisition window and hence contribute to the acquired image. For example, stimulated echoes can give rise to inverted ghosts overlying the main image.

Elimination of spurious echoes and the resulting artefacts requires careful pulse sequence design, particularly in terms of the phase encoding gradients since magnetisation retains memory of previous phase encoding episodes while stored along the longitudinal axis and this may later contribute to stimulated echo artefacts. Spoiler gradients may also need to be added to eliminate unwanted coherences, as is done routinely in the STEAM sequence.

Related Articles: Gradient echo (GE), Spin echo, Spoiling, STEAM (stimulated echo acquisition mode), Stimulated echo

Square wave oscillation

(General) See Saw-tooth voltage

Spurious coincidence

(Nuclear Medicine) This refers to the coincidence between either a $\gamma$-photon and an annihilation photon or two $\gamma$-photons. Spurious coincidences occur when imaging radionuclides which emit both positrons and high energy prompt cascade $\gamma$-rays. Such radionuclides are referred to as dirty radionuclides. These $\gamma$-photons are seldom coplanar or have a 180° correlation to the annihilation photon, hence leading to a misplaced line of response. These events can be falsely interpreted as true coincidences, that is provide false spatial information which degrades the spatial resolution. A PET system operating in 3D mode is more sensitive to spurious coincidences than while running in a 2D mode.

One of the two annihilation photons in Figure 5.90 is attenuated in the object while the other is detected. The dashed arrow represents the $\gamma$-photon direction. If the $\gamma$-photon is detected the line of response will be misplaced.

Spurious effects can be avoided by using a narrow energy window because the $\gamma$-photon energy often differs from 511 keV. The energy window in PET is typically set to 300–700 keV and events of an energy outside that window are discriminated. High-energy $\gamma$-photons (over 700 keV) can Compton scatter prior to detection and therefore avoid being discriminated by the energy thresholds. However a narrow energy window will also result in a low sensitivity.

Related Articles: Dirty radionuclides, PET

Sr-89-strontium chloride [Metastron™]

(Nuclear Medicine) Strontium-89 chloride, commonly known as Metastron™ (GE Healthcare) is a bone-seeking radiopharmaceutical used for pain palliation and is considered a clinically effective and cost-effective treatment in patients with advanced cancer metastatic to bone. Patients can often obtain pain relief for up to 6 months after a single injection of $^{89}$Sr-chloride, resulting in a significant improvement in quality of life.

$^{89}$Sr is a beta-emitter with maximum energy of 1495 keV and a maximum range of the beta particles in tissue of 8 mm. The half-life of $^{89}$Sr is 50.53 days. The $^{89}$Sr-chloride activity recommended for therapy of bone pain is 1.5–2.22 MBq kg$^{-1}$ with mean activity of 150 MBq given intravenously.

After an IV injection of $^{89}$Sr-SrCl$_2$ it localises in reactive bone and 80% is excreted in the urine and 20% with feces with a biological half-life of 4–5 days. It is assumed that approximately 30–35% of the administered activity retains in normal bone for 10–14 days. However, it has been reported that the retention could be as high as 85%–90% in osteoblastic areas 3 month after injection. Initial pain relief may be noticed already after 3–5 days, and the mean duration is of the order of 3–6 months.

The mean absorbed dose to vertebral metastasis from a IV injection of $^{89}$Sr-chloride has as an average value been estimated to be 230 mGy MBq$^{-1}$ (interval 60–610 mGy MBq$^{-1}$). The absorbed dose is 17 mGy MBq$^{-1}$ to bone surfaces and 11 mGy MBq$^{-1}$ to the red bone marrow. Other organs receive less than 5 mGy MBq$^{-1}$. The effective dose for $^{89}$Sr-chloride is approximately 2.9 mSv MBq$^{-1}$.

Related Articles: Sm-153-EDTMP [Lexidrom], Rhenium-186-hydroxyethylidene diphosphonate

SSFP
(Magnetic Resonance) See Steady state free precession (SSFP)

Stabilisation
(General) In radiology this term refers to the process of or stabilising or regulating the output of a power supply which can be subjected to changes in both input supply voltage and output power demand.

In order to provide proper regulation of the output of x-ray generators and other high-power electrical devices, it is necessary to stabilise the mains supply power to these devices.

Two techniques are used – the constant voltage transformer (CVT) and the automatic voltage stabiliser (AVS). Both are based around the use of a power transformer, and so are usually bulky devices.

The constant voltage transformer has a special ferromagnetic core which when operating normally is fully 'saturated' each half cycle by the induced field of the primary or input winding. Thus the output winding receives a magnetic field of constant maximum value independent of the input supply and therefore provides an output as though from an unfluctuating source. This type of transformer is good at absorbing large input voltage spikes, but is not as efficient as the AVS.

The automatic voltage stabiliser is a variable (auto) transformer controlled electronically or electromechanically to maintain a constant output voltage. The transformer output is converted to a DC value and compared to a preset reference voltage. When the output sags or rises due to a line or load change, the number of turns in one transformer winding are altered to compensate – either by a motor-driven contactor or by electronic semiconductor switching. This is a much more power efficient technique but may not be able to react instantaneously to changes, and will not be able to absorb large input voltage spikes.

Related Articles: Stabilisation, Variable transformer

Stabilised amorphous selenium (a-Se)
(Diagnostic Radiology) Selenium (Se) is a non-metal chemical element with an atomic number (Z) of 34. Stabilised amorphous selenium (stabilised a-Se) is the most widely used photoconductor in medical imaging. Amorphous selenium is stabilised by alloying it with 0.2%–0.5% arsenic (As) and doping with chlorine (Cl) in 10–20ppm range. The arsenic is introduced to prevent the structure from re-crystallizing while the Cl compensates for the hole traps introduced by the As. The amorphous nature of selenium is advantageous in the manufacture of flat panel detectors as it can be easily deposited on a suitable substrate by conventional vacuum deposition techniques to form large area photoconductive film of thicknesses up to 1000μm.

Related Articles: Amorphous selenium, a-Se photoconductive layer, Selenium detector, TFT (thin-film technology).


Stabiliser
(General) See Stabilisation

FIGURE S.91 Schematic diagram showing the relative number of protons and neutrons of stable nuclides.

Stable nuclei
(General) Stable nuclei are nuclei one that require the addition of energy to transform them into other nuclei. Stable nuclei do not spontaneously decay and can only be transformed by being bombarded by particles or photons under specific conditions. All radioactive decay processes eventually result in a stable nuclide, although there may be a number of intermediate radionuclides.

Stable nuclides with low mass (apart from hydrogen) have similar number of neutrons and protons but with increasing atomic number there is an increasing excess of neutrons over neutrons and the ‘line of nuclear stability’ deviates from the $N = Z$ line (Figure S.91). The reason for this lies in the characteristics of the nuclear force and the electrostatic repulsion between protons. The nuclear force has a very short range, less than that of a large nucleus, so additional neutrons are required in large nuclei to overcome the electrostatic repulsion between the protons.

A nuclide which does not have a stable combination of neutrons and protons (nucleons) transforms spontaneously into a stable (or more stable) nuclide through the process of radioactive decay.

Related Articles: Nuclear force, Nuclear instability, Nucleons, Nuclide, Radioactive decay

Standard man
(Radiation Protection) ‘Standard man’ was originally presented in 1949 at the Chalk River Conference on Permissible Dose. In 1963 the International Commission on Radiological Protection (ICRP) requested the revision and extension of standard man and recommended that the name was changed to ‘reference man’.

The need to establish a reference man arose due to a requirement to

a. Represent a typical radiation worker
b. Indicate some of the factors such as age and sex on which estimating dose to individuals depends

The full specification of reference man is given in ICRP Publication 23, Report of the Task Group on Reference Man (ICRP 1974). Many characteristics are defined in this publication including weight, length and surface area of the whole body for males and females of different ages.
Related Article: International Commission on Radiation Protection

Standard uptake value (SUV)
(Nuclear Medicine) The standard uptake value (SUV) is the ratio of the radionuclide concentration in the volume of interest to the mean radionuclide concentration throughout the body.

Standby position
(Diagnostic Radiology) See Parking position

Standing wave
(Radiotherapy) See Wave guide

Stannous chloride (Sn2+Cl2)
(Nuclear Medicine) Stannous chloride, Sn2+ Cl2, is the preferred reducing agent for 99mTc--radiopharmaceuticals that need to be labelled by a simple procedure shortly before use. The 99mTc--pertechnetate ion (NO3)99mTcO4 is in the oxidation state +VII and must be reduced to +I, +III, +IV or +V in order to give 99mTc--complexes with high labelling efficiency, shelf-life and purity.

Prefabricated 99mTc--kits offering labelling in isotonic solutions simply by adding pertechnetate in the evacuated vial. Stannous salts are reliable reducing agent used in all kit formations. SnCl is non-toxic and stable during lyophilisation (i.e. freeze drying).

Tin forms compounds in the oxidation states +II and +IV. The Tc-chemistry is difficult and not known in detail, but the first step is the reduction of Tc to (+V) from Tc (+VII), followed by two successive complementary reactions:

\[
\begin{align*}
\text{Tc(VII)} + \text{Sn(II)} & \rightarrow \text{Tc(V)} + \text{Sn (IV)} \\
\text{Tc(V)} + \text{Sn(II)} & \rightarrow \text{Tc(III)} + \text{Sn(IV)}
\end{align*}
\]

The reduced technetium is chemical reactive and can be labelled to different chemical molecules, for example chelates:

Reduced 99mTc + chelating substance \(\rightarrow\) 99mTc-- chelate.

Although Sn is the reducing agent of choice there are some problems. The chemistry of stannous compounds is complex and the Sn is difficult to purify. Thus, the labelled product contains Sn(IV). Sn is easily oxidised to Sn(IV) by the oxygen in air. The shelf-life of the preparation is dependent on the Sn concentration. The amount of Sn2+ is optimised for each pre-fabricated Tc-kit, in order to maintain that the amount is large enough to ensure reduction of both 99mTc and its daughter 99Tc, and conversely to be low enough in order to avoid further reduction of the 99mTcO4 to a lower oxidation state. The SnCl amount in a pre-fabricated 99mTc--kit ranges from 0.76 to 500 μg, corresponding to a Sn/Tc – ratio 10^4–10^8.

Since oxygen for most of the 99mTc--radiopharmaceutical kits may oxidise the Sn ions a air-valve cannula should not be used during the labelling.

Related Article: Tc-99m – sodium pertechnetate

Static electricity
(General) Static electricity refers to the accumulating of electric charges on the surfaces of objects. The static charges remain on an object until they either bleed off to ground or are rapidly neutralised by an electrostatic discharge.
The phenomenon of static electricity requires a separation of positive and negative charges of the contacting objects. Electrons can be exchanged between materials on contact; materials with weakly bound electrons tend to lose them, while materials with sparsely filled outer shells tend to gain them. This is known as the triboelectric effect and results in one material becoming positively charged and the other negatively charged. Separation of charge within the conductor can occur when a charged object is brought into the vicinity of an electrically neutral object. Charges of the same polarity are repelled and charges of the opposite polarity are attracted.

In some materials separation of charges can be generated also by mechanical pressure (piezoelectric materials) or by heating (pyroelectric materials).

The feeling of a static electric shock is caused by the stimulation of nerves as the electrostatic discharge current flows through the human body. Usually the accumulated charge is generally not enough to cause a hazardously high current. The spark associated with static electricity is caused by electrostatic discharge as excess charge is neutralised by a flow of charges from or to the surroundings.

Despite the apparently harmless nature of static electricity as we generally experience it, there can be significant risks where large charges may accumulate in the presence of sensitive materials or devices.

**Related Articles:** Spontaneous discharge, Voltage limiter

**Further Reading:** [http://en.wikipedia.org/wiki/Static_electricity](http://en.wikipedia.org/wiki/Static_electricity)

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**Static field**

*Magnetic Resonance*

**Introduction:** Magnetic resonance imaging makes use of three types of magnetic field: a static field, a pulsed field varying linearly in space (the imaging gradients) and a radiofrequency (RF) field. Of these, the static field is the most conspicuous, and arguably of overriding importance in determining the quality of images obtained and the range of techniques that can be performed on a given MR imaging system. It is also the most problematic from a safety perspective.

The purpose of the static magnetic field is to cause magnetic nuclei in the imaged object to align with a component of their angular momentum either parallel or antiparallel to the field direction, corresponding to two different energy levels and hence endowing the object as a whole with a small net longitudinal magnetisation because of the difference in population of these two levels determined by the Boltzmann distribution. Because the net magnetisation is proportional to the strength of the static field, and because it is this magnetisation that determines the amount of signal available for image formation and hence is the limiting factor for image resolution in space and time, much effort has been expended in the development of whole-body MRI devices with ever-increasing static field strengths. This requirement, in turn, has been a major driver in the development of superconducting magnet technology.

**Historical Aspects:** Early MRI researchers made use of resistive electromagnets with field strengths of up to about 0.15 T. At higher field strengths, these magnets required water cooling to dissipate heat generated by the considerable electrical currents flowing through the magnet windings. The development of whole-body superconducting magnets in the mid-1980s allowed construction of scanners with static field strengths of up to 1.5 T or more, without resistive heating. However, these systems introduced the new problem of fringe fields extending to a considerable distance from the magnet with concomitant difficulties in siting (either because of the field itself or the tonnes of steel shielding required to contain it). Introduction of active shielding did much to alleviate this situation, but arguably has exacerbated the safety problems around static fields, in that it introduces a very steep static field gradient close to the scanner with little prior warning.

**The Market Today:** Today, whole body superconducting magnets (usually based on niobium-titanium (NbTi) wire) are in use with field strengths of up to 9.4 T, with yet higher field systems under rapid development. Scanners at 7 T are becoming an established feature of major MRI research centres, while 3 T is almost commonplace. Ultrahigh field machines have undoubted benefits in areas such as functional MRI and perfusion imaging, as well as opening up new contrast mechanisms for exploitation. However, these benefits are offset by a worsening of problems such as magnetic field inhomogeneity, susceptibility artefacts, RF penetration and specific absorption rate (SAR).

There is certainly still a place in the market for lower field scanners, today usually based around permanent magnets, which have lower operating costs than resistive electromagnets. This is particularly so for open architecture scanners, which are better suited to obese and paediatric patient populations and for many types of interventional MRI.

**Performance Specification:** The most important performance parameter related to the static field, apart from the field strength itself, is the uniformity of the field. This determines the useful field of view (FOV) of the scanner, and is often specified in terms of the field variation (in ppm) over a spherical volume of a specific diameter (DSV) at the scanner isocentre. For example, a modern 1.5 T superconducting system might have field homogeneity of 0.2–0.4 ppm over a 40 cm DSV, while for a lower field open system this might be 2–4 ppm over a 20 cm DSV.

From a practical point of view, the extent of the fringe field will have a major impact on ease of siting and operation, and potentially on interference with other activities in the vicinity. The 0.5 mT isoco- contour is often taken to define the extent of the fringe field for safety purposes. However, it is important to recognise that some devices, such as PET scanners, may be affected at significantly lower field strengths.

**Safety Aspects:** The static field presents the greatest safety challenge in MRI, since ferromagnetic objects brought close to the magnet are subject to considerable force and torque and can be turned into high-speed projectiles. Death and serious injury have resulted from accidents of this sort. There is also a risk that the field may cause malfunction of implanted active medical devices (IAMDs) or injury due to the force exerted on IAMDs or ferrous foreign bodies such as shrapnel. Most safety precautions in place in MRI units are designed to ameliorate this hazard.

In terms of direct biological effects, the phenomenon most frequently reported is vertigo and related symptoms when moving the head in a strong static field, usually 3 T or above. These transient effects are increasingly well understood. Measurable changes in blood pressure have occasionally been reported, probably related to induction of electrical potentials due to movement of charged ions in blood through the magnetic field. These are physiologically negligible at least up to 8 T.

**Related Articles:** Boltzmann distribution, Longitudinal magnetisation, Magnet, Magnet(s) superconducting, Nuclear magnetic resonance (NMR), Tesla

---

**Stationary anode**

*Diagnostic Radiology* X-ray tubes with stationary anode are used for low power (usually up to 2 kW) x-ray equipment – for example...
some small mobile and dental x-ray equipment (Figure S.93). The simplest construction of stationary anode is a large copper stem with a small tungsten plate (2–3 mm thick) embedded at its target side (exactly opposite the cathode). The thermal conductivity of the copper is sufficient to remove the heat from the target to the surrounding space. For tubes used for frequent exposures the copper end of the anode can be cooled with non-conducting mineral oil (see the article Anode).

Some undesirable effects happen during the bombardment of the anode. First, some of the bombarding electrons are elastically scattered backwards from the target, enter again the accelerating field and fall again to the anode, but this time outside their original target place. This creates extrafocal radiation (generated outside the focal spot), also called off-focus radiation or stem radiation. This off-focus radiation effectively enlarges the focal spot and leads to blurred image. The scattered electrons can also hit the glass, what can not only metalise it but even rupture it and affect the vacuum in it. A metal ‘electron-capture’ hood is placed over the anode in order to retain these electrons. The anode hood is made of copper with an opening at the side of the cathode. At the path of the useful radiation the copper is replaced with Beryllium plate which can pass the x-rays with minimal absorption ($Z = 4$ for Be). Beryllium ‘window’ is widely used in many contemporary x-ray tubes (Figure S.94).

Because historically the first x-ray tubes have been with stationary anode we can use them for easier explanation of the Line focus principle. In the early x-ray tubes the angle between the target plate and the electron beam has been 45°. In 1918 German scientist Goetze has patented a tube with much smaller anode angle $\alpha$, thus increasing the area bombarded by the electrons—the actual focus $AF$ (or the thermal focus, $F_t$, also called actual focus $F_a$) and decreasing the size of the effective focus $EF$ ($F_e$, also called optical focus). This allowed increasing of x-ray tube power (due to increasing the anode heat capacity) and increasing the sharpness of x-ray image (the smaller optical focus acts as smaller point source of ‘light’). In fact this new design affected the length of the actual focus (proportional to the length of the filament spiral coil) while the width of the actual focus (equal to the width of the filament spiral coil) remains unchanged. Due to this reason the principle for enlarging the area of the actual focus, while keeping the effective focus small is called Line focus principle (or Goetze principle). From Figure S.95 one can see that the relation between the actual focus and the effective focus depends on the sine of the anode angle:

$$F_e = \sin \alpha \cdot F_a$$

There are two stationary anodes on Figure S.95 – with anode angle $\alpha$ of 20° (Figure S.95a) and with 10° anode (Figure S.95b). On the diagram, $a$ is the projection of the cathode filament over the anode target. It is obvious that the size of $F_e$ corresponds to the width of the filament spiral coil. Both x-ray tubes have similar effective focus ($F_e$) but the tube with smaller anode angle (Figure S.95b) has larger actual focus ($F_a$) and this way the heat is distributed over larger area of the target.

The anode angle of most contemporary diagnostic x-ray tubes varies between 10° and 20° and for the therapeutic x-ray tubes it is still 45°. The increase of the thermal focus allowed better heat distribution over the anode target, but the need for further increase
the x-ray tube power lead to the development of the x-ray tubes with rotating anode.

**Related Articles:** Anode, Rotating anode, Target

**Hyperlink:** EMERALD (DR module), www.emerald2.eu

### Stationary grid

*(Diagnostic Radiology)* A stationary grid is not moved (as in a Bucky device) during an x-ray exposure. Often it leaves over the film undesired image of the grid strips/lines.

**Related Articles:** Grid, Bucky

### Stator

*(Diagnostic Radiology)* The x-ray tubes with rotating anode use an induction electrical motor. The copper rotor of this motor is inside the glass envelope of the tube and rotates the anode disk. The stator consists of a number of electromagnets (solenoids) placed outside the glass envelope. These are switched on and off in a rapid sequence, thus creating a rotating electromagnetic field, which drags the rotor to turn in the same direction. The mechanical speed of rotation of the rotor (rpm) can never reach the speed which drags the rotor to turn in the same direction. The mechanical slipping of the rotor (due to mechanical slipping).

All control of acceleration, speed of anode rotation and deceleration (stopping) of a rotating anode is done through variation of the stator electrical supply. See the model of a stator in the article Rotating anode.

**Related Articles:** Anode, Rotation anode, Anode rotation speed, Starting device, Bearing


### Steady state

*(Magnetic Resonance)* The term steady state is generally used to describe a situation where a given property of a system does not change over time. In MRI, steady state typically refers to a period of time during which the magnetisation is stable. For example, longitudinal steady state requires that the longitudinal magnetisation is unchanged over time while a transverse steady-state condition implies that the transverse magnetisation attains a non-zero steady state. In practice, the term steady state is typically used for the case when the transverse magnetisation at each radiofrequency (RF) pulse in a sequence, or at a given time after it, is the same as at the corresponding point at, or after, the next RF pulse in the sequence.

**Related Articles:** CISS constructive interference in the steady state, Dual echo steady state (DESS), Fast imaging with steady state precession (FISP), Spoiled gradient recalled acquisition in the steady state (SPGR), SSFP steady state free precession

### Steady state condition in tracer kinetic modelling

*(Nuclear Medicine)* A steady state system refers to a system in which a process, parameter or variable is not changing with time. A biochemical flux is said to be in a steady state when the concentrations of reactants and products are steady over time. Such a system seldom exists because different systems have different biorhythms, for example the vascular system flux varies with heartbeats. However if the experimental sampling rate is slow in comparison to the biorhythm, for example the sample is collected over a series of heartbeats, the sample represents the mean flux of blood, and thus the steady state assumption is valid. In an opposite situation where the experiment sample rate is faster than the biorhythm the sample seldom represents a steady state condition.

It is important not to confuse the steady state of a process with the steady state of a tracer. Measurements are often made while the tracer is distributing within the process under study. Tracer kinetic models that measure the desired function while both the tracer and process are in a steady state are usually referred to as equilibrium models.

**Related Articles:** Analogue tracer, Distribution volume, Partition coefficient


### Steady state free precession (SSFP)

*(Magnetic Resonance)* Steady state free precession refers to the use of steady state gradient echo sequences (GRE) using short repetition and short echo times. Due to the rapid repetition of excitation pulses, the intensity of the MR-signal (the amplitude of the magnetisation vector) will initially fluctuate before eventually reach a steady state in the longitudinal direction. For the very short repetition times even the $T_2^*$-decay is not complete at the end of each sequence repetition (i.e. at the next excitation pulse) and there remain a residual component of the transversal magnetisation vector before the next excitation. Part of this transversal component will be refocused by the next excitation rf-pulse. The overall transversal magnetisation is the sum of different transversal magnetisation components, which can cause coherently or destructively interferes. Therefore steady state gradient echo sequences are divided into two groups; steady state incoherent (SSI) sequences, where the remaining transverse magnetisation is spoiled, and steady state coherent (SSC) sequences, where the use of balanced gradients are introduced. The latter is accomplished through the use of rephasing/refocusing gradient pulses, added so that the gradient scheme becomes balanced over one repetition time, that is the gradient area is summing up to zero. Therefore the latter group is also denoted as balanced steady state free precession (bSSFP) sequences, which results in a high SNR.

The rapid repetition of excitation pulses in an SSC sequence will also lead to the generation of several different signal components since any rapid repetition of three or more RF pulses will in fact generate at least five different echo signals. Depending upon the timing of the SSFP pulse sequence, the MR signal will build from several different components and it is crucial to keep track of which signal is actually collected. The SSFP-FID signal is the gradient refocused echo which is collected immediately after an excitation pulse, whereas the SSFP-echo signal is generated after two consecutive RF pulses, where the second pulse is seen upon as a refocusing pulse (similar to the 180° in the SE case) for the spins which were excited by the preceding RF pulse. This leads to the somewhat confusing consequence that the SSFP-echo sequence has a TE which is longer than the TR. Since the SSFP-echo sequence collects the phase coherent signal just prior to an RF pulse, the pulse sequence structure is a time-reversed SSFP-FID sequence. The SSFP sequences are designed to collect signals which are generated through many different coherence pathways and the contrast is thus a combination of both $T_1$ and $T_2$ relaxation effects. A pulse sequence can be designed to look at either one (see FISP and PSIF), or both (see CISS and DESS), of the two signal components and depending upon which signal is detected different contrasts will be obtained.

**Related Articles:** CISS, Constructive interference in the steady state, Dual echo steady state (DESS), Fast imaging with steady state precession (FISP), Gradient echo (GRE), PSIF (FISP reversed), T2

**Further Readings:** Scheffler, K. 1999. A pictorial description of steady-states in rapid magnetic resonance imaging. *Concept.*
Steel is an alloy consisting predominantly of iron with 0.2%–2.14% carbon and other elements including chromium, manganese and tungsten. The purpose of the alloying elements is to improve the properties of steel, such as tensile strength and hardness, by preventing the movement of dislocations in the crystal lattice. However, as hardness increases the steel becomes more brittle and so a compromise is made.

Steel is classified by a number of grades defined by standards organisations. There are several types of steel, such as carbon steel, stainless steel (includes chromium to increase corrosion resistance) and maraging steel (includes nickel making it malleable). Iron alloys have various phases, such as austenite, cementite and cast iron, which has a lower melting point than steel and good castability.

Steel became common when efficient production methods were invented in the seventeenth century and became relatively cost-effective when the Bessemer process was devised in the nineteenth century. Today steel is widely used in construction, vehicles and many appliances.

**Medical Applications:** Because of its strength, steel is used extensively for medical applications from surgical instruments (naturally stainless steel to prevent corrosion) to components of medical equipment. Steel can be non-magnetic, making it suitable for use in MRI controlled areas. In magnetic form it is widely used in permanent magnetic cores used in MRI scanners. However superconducting magnets are most commonly used in modern machines, due to the higher magnetic field strengths achievable.

**Related Articles:** Magnetic resonance imaging, Controlled area

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**Step wedge**

*(Radiotherapy)* A step wedge was originally a concept used for testing the contrast and brightness performance of display monitors, and consists of a set of 11 squares ranging from 0% (black) to 100% (white) in steps of 10% (see Attenuation steps). An equivalent to this can be delivered on a linac (typically by delivering a series of fields by moving the MLC leaves) where the steps relate to different doses and this is used for calibration of film or some other dosimeter.

**Abbreviation:** MLC = Multi leaf collimator.

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**Step-down transformer**

*(Diagnostic Radiology)* Transformer producing lower output voltage.

See article on Transformer

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**Step-up transformer**

*(Diagnostic Radiology)* Transformer producing higher output voltage.

See article on Transformer

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**Stereotactic frame**

*(Radiotherapy)* A stereotactic frame is used to enable highly accurate patient set-up for stereotactic radiosurgery. Such a device is often used in conjunction with a gamma knife or a conventional linear accelerator. The frame is attached to the patient’s head, often using bite blocks and screw fixation. The frame generally has a pattern of radio-opaque markers to enable localisation during CT imaging to plan the treatment and/or contrast markers to enable location of the frame in an MR scan. Two approaches to attaching the frame to the treatment room are used: pedestal mounting and couch mounting. A common example is the Gill–Thomas–Cosman frame.

**Abbreviations:** CT = Computed tomography and MR = Magnetic resonance.

**Related Articles:** Gamma knife, Robotic linac, Arc therapy, Stereotactic radiosurgery


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**Stereotactic radiosurgery**

*(Radiotherapy)* Stereotactic radiosurgery is a radiotherapy technique for delivering treatments with high spatial accuracy, particularly to the brain. It does not involve traditional surgery contrary to its name.

It is used to treat both malignant and benign tumours. It is also used to treat arteriovenous malformations (AVMs). Often stereotactic radiosurgery is delivered as a single fraction treatment, although it may also be fractionated.

Technology for stereotactic radiosurgery falls into three classes: the gamma knife, robotic linacs (such as the Cyberknife) and conventional linear accelerators. Treatment with conventional linear accelerators often involves the use of several arcs to focus the dose distribution onto the target. The patient is often located in a stereotactic frame to improve set-up accuracy.

**Abbreviation:** AVM = Arteriovenous malformation.

**Related Articles:** Gamma knife, Robotic linac, Arc therapy, Stereotactic frame

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**Stimulated acoustic emission**

*(Ultrasound)* This is a phenomenon discovered during the early clinical use of contrast agents. After a period of not transmitting ultrasound (i.e. the system in ‘freeze’-mode) a short-lived ‘mosaic’ of colours would be seen in colour Doppler mode, as the system was activated. The phenomenon was explained as a result of rupturing contrast agent particles (microbubbles) that caused a random
pseudo Doppler shift: microbubbles with a stabilizing shell, as used in contrast agents, behave differently in an acoustic field, depending on the transmitted power. At low emission power, the bubbles act as linear backscatters, but with increasing power, they start to respond non-linearly, which result in harmonic frequencies. At high power levels as used in colour flow imaging, they disintegrate, and the free gas that is released is rapidly dissolved. As colour Doppler systems detect if there has been a phase shift between consecutive pulses, the sudden disappearance of an echo will be interpreted as a phase shift of random magnitude. In fact the phase shift is not entirely random, but will on average have a slight negative bias, depending on acoustic power and microbubble concentration. The phenomenon has been suggested for clinical use in a more formal way according to a scheme called sono-scintigraphy.

**Stimulated echo**  
(Magnetic Resonance) A stimulated echo is a type of spin echo signal produced by a series of three RF pulses rather than from the combination of a 90° and 180° pulse as in a conventional spin echo sequence.

The process by which a stimulated echo is formed is illustrated below for the case of a train of three 90° pulses, although in general any train of pulses with flip angles other than 180° can produce a stimulated echo (Figure S.96).

The initial RF pulse generates transverse magnetisation. It is assumed in this example that this magnetisation is fully dephased by the time the second RF pulse is applied (a spoiler gradient can be employed to ensure that this is the case), so that transverse magnetisation is distributed isotropically in the x–y plane before the second RF pulse is applied. The y-component of each magnetisation isochromat will be rotated to the z-axis by this second pulse. The longitudinal magnetisation generated in this way is ‘stored’ along the z-axis until application of the final pulse, while remaining transverse magnetisation dephases (again with the aid of a spoiler gradient if needed). Application of the third RF pulse returns stored magnetisation to the transverse plane where it refocuses as a so-called stimulated echo. The individual pulses generate slice-selective free induction decay (FID) signals, and each pair of pulses a two-dimensionally selective echo; these unwanted signals are dephased using spoiler gradients (see Figure S.97).

The use of a stimulated echo, rather than a spin echo as in PRESS, results in loss of 50% of the available signal. The effects

**Stimulated echo acquisition mode (STEAM)**  
(Magnetic Resonance) STEAM is a common spatial localisation technique used for single voxel spectroscopy (SVS). Because it involves acquisition of a stimulated echo signal some time after excitation, it is particularly suited to nuclear species with long T2 relaxation times, such as hydrogen (1H) nuclei (protons).

STEAM is a ‘single shot’ technique, in that it requires a single acquisition of the pulse sequence to achieve localisation, and is not dependent on post-acquisition signal combination (as is the case with, e.g. ISIS).

The STEAM pulse sequence uses three selective 90° pulses in turn, each applied with a gradient along a different Cartesian axis, and hence selecting orthogonal planes of spins (Figure S.97). The first pulse and gradient combination is used to excite a plane of spins, and when the second is applied, spins at the intersection of the two selected slices are returned to the z-axis during the ‘mixing time’ T_M, when the magnetisation experiences T1-relaxation. The third pulse brings magnetisation within the volume of interest at the intersection of all three slices back into the transverse plane, where it forms a so-called stimulated echo. The individual pulses generate slice-selective free induction decay (FID) signals, and each pair of pulses a two-dimensionally selective echo; these unwanted signals are dephased using spoiler gradients (see Figure S.97).

The use of a stimulated echo, rather than a spin echo as in PRESS, results in loss of 50% of the available signal. The effects

![Formation of a stimulated echo](image1)

**FIGURE S.96** Formation of a stimulated echo.

![STEAM pulse sequence](image2)

**FIGURE S.97** STEAM pulse sequence. (From Keevil, S.F., Phys. Med. Biol., 51, R579, 2006.)
of J-modulation can be minimised in STEAM by using a short echo time \( (T_E) \). STEAM is a popular technique because the VOI is well defined and contamination with extraneous signal is minimal.

**Related Articles:** ISIS, PRESS, Magnetic coupling, Magnetic resonance spectroscopy, Mixing time, Single voxel spectroscopy


**STIR** (short TI/tau inversion recovery)

(Magnetic Resonance) See Short tau inversion recovery (STIR)

**Stochastic effects**

(Radiation Protection) There are two types of biological effects (Bioeffects) of ionising radiation on human tissues categorised by the risk of the effects being observed. These two categories are stochastic and non-stochastic or deterministic effects.

Effects, which are statistically detectable only in populations are termed ‘stochastic effects’ because of their random nature. Stochastic effects may occur if an irradiated cell is modified rather than killed.

Modified abnormal cells may, after a prolonged process, develop into a cancer. Because the body’s defences are not completely effective, the smallest number of modified cells may develop into a cancer and there is assumed to be no threshold below which effects do not occur. Furthermore, since the probability of developing a cancer is partly dependent on the initial number of modified cells, these so-called stochastic effects occur with the probability rather than the severity of the effect being a function of radiation dose.

If the cell damaged by radiation exposure is a germ cell, whose function is to transmit genetic information to progeny, it is conceivable that hereditable effects of various types may develop in the descendants of the exposed individual.

As shown on Figure S.98, the current internationally accepted framework for radiation protection assumes a linear relationship between the risk (probability) of a stochastic effect, and the dose received, with no threshold – that is the linear non-threshold model.

Stochastic effects are restricted to cancers and hereditary disease only and are normally expressed as a risk of the effect occurring per unit dose. Current values taken from ICRP 103 (2007) are shown in Table S.3.

It is important to note that malignancies induced as a result of radiation damage are not distinguishable from those induced by other causes. It must also be remembered that the time interval between exposure and the manifestation of the leukaemia or solid tumour – the so-called latent period – may be tens of years:

- Leukaemia: latent period of 5–10 years
- Solid cancers: latent period 20–30 years

**Related Articles:** Bioeffects, LNT Model


**Stopping power**

(Radiation Protection) The energy lost from a beam of charged particle (e.g. alpha or beta) ionising radiation as it travels through a medium is known as the stopping power of the material traversed. This loss of energy for alpha particles, and protons (but not electrons) is described by the Bethe–Bloch Equation.

The stopping power can be described in terms of distance travelled through the medium (linear stopping power), or in terms of the atomic mass of the medium (collisional mass stopping power).

The term ‘stopping power’ is used to imply that the particles will have a finite range in matter whereby all incident and secondary particles have been totally absorbed within the medium (as opposed to x- or \( \gamma \)-radiation that is attenuated exponentially with at least some transmission through the material and out the other side).

**Related Articles:** Linear stopping power, Collision mass stopping power, Collisional energy loss, Bethe–Bloch equation

**Stopping power**

(Radiotherapy) As a fast charged particle passes through matter it interacts with elastic and inelastic collision with atomic electrons and nuclei. The various types of interactions produce a loss of energy bringing the charged particle eventually to rest. The evaluation of the energy loss should include energy losses due to all types of interactions but the predominant mode of energy loss is the ionisation of the medium atoms. The average linear rate of energy loss per unit of path length \( x \) by a charged particle with a kinetic energy \( E \) in a medium of atomic number \( Z \) is called stopping power. Common units for the stopping power are MeV cm\(^{-1}\). Using the relativistic quantum mechanics the following expression for the stopping power can be derived:

\[
\frac{dE}{dx} = \frac{4\pi k_0 z^2 e^4 n}{mc^2 \beta^2} \left[ \ln \left( 2mc^2 \beta^2 I(1-\beta^2) - \beta^2 \right) \right]
\]

where
\( k_0 \) is the 8.99 10\(^9\) N m\(^2\) C\(^{-2}\)
\( z \) is the atomic number of the heavy particle
\( e \) is the magnitude of the electron charge
\( n \) is the number of electrons per unit volume in the medium
\( m \) is the electron rest mass
\( c \) is the speed of light in vacuum
\( \beta = v/c \) is the speed of the particle relative to \( c \)
\( I \) is the mean excitation energy of the medium

---

**Table S.3**

<table>
<thead>
<tr>
<th>ICRP Detriment-Adjusted Nominal Risk Coefficients (ICRP 103) (10-2 Sv-1 – Percent per Sievert)</th>
<th>Exposed Population</th>
<th>Cancer Induction</th>
<th>Hereditable Effects</th>
</tr>
</thead>
<tbody>
<tr>
<td>Whole</td>
<td>5.5</td>
<td>0.2</td>
<td></td>
</tr>
<tr>
<td>Adult</td>
<td>4.1</td>
<td>0.1</td>
<td></td>
</tr>
</tbody>
</table>

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**Figure S.98** Dose–response curve for stochastic effects.
At low energies ($\beta \to 0$) the factor in front of the bracket increases while the logarithmic term increases the stopping power at very high energies ($\beta \to 1$). Nevertheless the logarithm term then decreases causing a peak called Bragg peak where the linear rate of energy loss is a maximum.

In the cavity theory the stopping power ratio, a dimensionless quantity, is used to describe the rate of energy loss of the charged particles in one medium in relation with another.

When a charged particle suffers an acceleration it radiates electromagnetic energy and the intensity of the emitted radiation is proportional to the acceleration times the charge. For a particle of mass $M$ and charge $z$ being accelerated by a charge $Ze$ the acceleration is proportional to $Ze^2/M$ and the intensity of the emitted radiation is therefore proportional to $(Ze^2/M)^2$. A light particle such as an electron is therefore more efficient to produce bremsstrahlung than a heavy particle of the same energy.

The stopping power could be subdivided into a collision stopping power $S_{\text{col}}$ and a radiative stopping power $S_{\text{rad}}$. The collision stopping power, due to inelastic collisions with atomic electrons of the medium, results in excitation and ionisations. The radiative stopping power, due to the particle interaction with the electric field of the nucleus, results in the production of bremsstrahlung radiation. The total stopping power is the sum of the two given by the following equation:

$$ \frac{dE}{dx} = S_{\text{col}} + S_{\text{rad}} $$

$$ \frac{dE}{dx} = \left( \frac{dE}{dx} \right)_{\text{col}} + \left( \frac{dE}{dx} \right)_{\text{rad}} $$

This separation emphasises the difference between the two components as the energy lost in ionisations and excitations is absorbed close to the charge particle track while the energy carried as form of bremsstrahlung travels far before being absorbed.

The collisional stopping power for electrons is different from that of heavy charged particles because an electron can lose a large fraction of its energy in a single collision with an atomic electron as it has the same mass and also because the two electrons are identical and in quantum mechanics the identity of particles implies that they cannot be distinguish.

The collision stopping power for electron can be written as

$$ \frac{dE}{dx} = \frac{4 \pi k_0^2 z^2 e^4 n}{mc^2 \beta^2} \ln \left( \frac{mc^2 \beta \sqrt{\beta^2 + 2}}{\sqrt{2I}} \right) + F(\beta) - \delta $$

where

$\tau$ is the $E_{\text{kin}}c^2$ with $E$ kinetic energy of the electron

$\delta$ is the density effect correction

$$ F(\beta) = \frac{1 - \beta^2}{2} \left[ 1 + \frac{\tau^2}{8} - (2\tau + 1)\ln 2 \right] $$

The following approximate formula gives the ratio of the radiative and collision stopping power for an electron:

$$ \frac{(-dE/dx)_{\text{rad}}}{(-dE/dx)_{\text{col}}} \approx \frac{(Z + 1.2)E}{800} $$

where

$E$ is the electron total energy expressed in MeV

$Z$ is the atomic number of the target material

**Storage phosphor**

*(Diagnostic Radiology)* Storage phosphors are phosphors that are designed to trap x-ray energy until the energy is released by laser readout. They are the basis for computed radiography systems.

These phosphors were patented by Kodak but the first commercial systems were introduced by Fuji. The storage phosphor is usually made from BaFX:Eu$^{2+}$ ($X = \text{Cl, Br, I}$). It is a thin flat sheet contained within a cassette, similar in appearance to the x-ray cassettes used in film-screen radiography.

The above drawing (Figure S.99) shows the basis of a storage phosphor.

---

**FIGURE S.99** Principle of photostimulated luminescence. (Drawing courtesy of JA Seibert.)
The incident x-ray energy moves an electron from the Eu$^{2+}$, converting it to Eu$^{3+}$ + free electron. This electron is captured by the bromine energy traps.

The number of produced free electrons is proportional to the energy and intensity of the incident x-ray beam. This way the areas of the storage phosphor, where electrons are captured, and the number of these electrons, correspond to the area of the x-ray fields with varying intensity – that is a latent image of trapped electrons is formed.

In the absence of other external energy (e.g. heat) the captures electrons can stay long time into these f-trapping centres – that is the latent image is stored in the phosphor.

When the storage phosphor is scanned with external laser (usually Infrared, e.g. He-Ne 632 nm) the trapped electrons gain sufficient energy to be move out of the trap. The released electrons during this process (reading) return to the Europium. During this process it returns back to its original Eu$^{2+}$ state and releases characteristic quant with 390 nm (photostimulated luminescence, PSL).

The above drawing (Figure S.100) shows the process of reading the storage phosphor. A reading laser scans the whole storage phosphor plate and the released PSL is read by a photomultiplier (through a light channeling guide). The intensity of the read light is proportional to the number of released electrons (from the traps) and hence to the intensity of the x-ray beam radiated the scanned area. As the laser and PSL lights are with different wavelength the reading laser does not interfere with the PSL light.

The read information from the phosphor plate enters the image processor, where it is arranged in a digital image matrix (thus forming the visual image of the radiograph).

After the reading process the storage phosphor plates are erased by exposing them to a bright light. This way the storage phosphor returns into its original state and is ready to be used again. The number of uses of a storage phosphor is more than 10,000 (Figure S.101 and S.102).

**Strain imaging**

(Ultrasound) Strain imaging assesses the tissue stiffness by compressing the tissue (with the ultrasound probe) and obtaining the local tissue deformation, which translates to percent shortening in the longitudinal dimension (assessed from apical views) or percent thickening in the radial dimension from a sequence of images. The technique provides a 2D representation of the tissue stiffness and facilitates visualisation of deep lying tumors, which is difficult with palpation alone. The technique has been developed into ultrasound elastography.

In cardiology the use of strain imaging combined with routine B-mode and M-mode assessment increases sensitivity compared with the use of visual assessment alone for determination of myocardial viability or regional myocardial function.

**Related Article:** Elastography

**Straton**

(Diagnostic Radiology) Straton is a vendor name (SIEMENS) of a specific x-ray tube used in computed tomography. The tube uses special focusing and deflection coils to form the beam of thermal electron bombarding the anode. This way the shape of this metal x-ray tube is unusual and allows variable focal spot. The stationary round anode of Straton is sealed with the tube metal envelope and...
rotates with it. This design allows for the back side of anode to be in direct contact with the cooling oil, hence increased power of the x-ray tube (see the article CT x-ray tube).

Related Article: CT x-ray tube

Stray magnetic field
(Magnetic Resonance) The stray field, called also fringe field, is the magnetic field outside the magnet of a MR system. Its strength depends on the magnet type and the magnetic field strength. The higher the field strength, the larger the fringe field. The stray field can be reduced through the use of a high permeability material that provides a return path of the magnetic flux with a significant decrease in the flux away from the magnet (passively shielded magnets). An alternative and more diffused method to control the stray field is to use an active shielding which consists in adding appropriate additional superconducting coils to superconducting magnets thus resulting in a significant reduction of the extent of the stray fields (actively shielded magnets). However the strength of the stray magnetic field of actively shielded magnets usually rises very rapidly. The stray magnetic field constitute one of the major hazards of MR scanners as these fields acting over extended distances outside the magnet produce strong attractive forces upon magnetic objects.

Streamline
(Ultrasound) Streamlines are curves that show the instantaneous direction of velocity of every particle along its line. There is no flow across the streamline. A bundle of streamlines enclosing a volume of fluid is known as a streamtube and no flow crosses this (Figure S.103).

A pathline shows the path of an individual particle over time.

Streaklines show the position of all particles that have passed a particular point; for example if dye is injected from a point, the resulting pattern is the streakline.

In constant laminar flow, the streamline, particle path and streakline coincide. For unsteady flow they differ.

The term streamline flow is used to describe steady flow without turbulence.

Related Article: Laminar flow

FIGURE S.102 Storage phosphor plate in a cassette.

FIGURE S.103 Streamlines and the enclosed streamtube.

Stress echocardiography
(Ultrasound) Stress echocardiography is echocardiography that it is performed both at rest and after increased cardiac work, or stress. This is invoked in the patient by exercise either on a bicycle or treadmill or by using a drug to increase heart rate. After a myocardial infarction some of the heart muscle contracts less efficiently. In coronary artery disease, narrowed arteries restrict blood flow to the heart muscle. These effects may be minimal or undetected at rest but become significant and evident when the heart performs more work – hence the clinical need to perform the investigation after exercise.

Related Article: Echocardiography

Stripping foil
(Nuclear Medicine) A stripping foil is used to convert H\(^+\) ions to protons by stripping it of two electrons in a negative ion cyclotron. In modern negative ion cyclotrons two foils are used to gradually extract the beam to two separate targets. When the H\(^+\) ion beam passes the foil the two electrons are stripped from the atom leaving nothing but a proton with an opposite charge. Since the proton has positive charge it will bend outwards where it can be focused on a target.

Related Article: Cyclotron

Structured noise
(Nuclear Medicine) Image noise can take the form of either random noise or structured noise. In nuclear medicine, random noise refers to the mottled appearance of the images due to the statistical nature of the acquisition. Structured noise refers to non-random variations in counts which interfere with the structures of interest within the image.

Structured noise can be caused by the distribution of the tracer. An example of this might be activity in the gut masking the myocardium in myocardial perfusion imaging. It can also be the result of imaging system problems such as artefacts caused by camera non-uniformity.

Related Article: Noise

Subject contrast
(Diagnostic Radiology) The formation of an x-ray image is the transfer and conversion of different types of contrast. If an object in the body is to be visible it must have some form of physical contrast in relation to its surrounding background. This can be in the form of a difference in density or atomic number (Z).

As an x-ray beam passes through the body section the object will attenuate either more or less than the surrounding tissue and will form an image where the contrast is in the form of exposure differences as shown in Figure S.104. The contrast in the x-ray beam coming from the patient’s body and exposing the receptor is designated as the subject contrast.

For an object that is being imaged the subject contrast can be expressed as the ratio of the difference in the object are exposure
**Subtraction** *(Diagnostic Radiology)* See Digital subtraction angiography *(DSA)*

**Summed scoring** *(Nuclear Medicine)* This is a scoring system used in the interpretation of myocardial perfusion SPECT scans. The myocardium is divided into 20 segments and each is given a score of 0–4 where 0 represents normal uptake and 4 no uptake at all. This 5-point scoring system leads to the derivation of three global summed scoring indices of perfusion. These are as follows:

1. The summed stress scoring SSS
2. The summed rest scoring SRS
3. The summed difference scoring SDS

The SSS and SRS are defined as the sum of stress scores for the 20 segments and the sum of rest scores for the 20 segments, respectively. Moreover, the SDS is defined as the difference between the SSS and SRS.

The SSS can be interpreted as follows:

- A SSS of less than 4 is considered normal or nearly normal
- A SSS of 4–8 is considered mildly abnormal
- A SSS of 9–13 is considered moderately abnormal
- A SSS greater than 13 is considered severely abnormal

**Related Article:** Scoring


**Superconducting magnet** *(Magnetic Resonance)* A superconducting magnet is a type of electromagnet that is built of superconducting coils and can reach much higher magnetic field strength than conventional electromagnet or permanent magnets. The coil is made up of tiny filament (~20μm) made of a type two superconductor (e.g. niobium-titanium, Nb,Ti). The type two superconductors become superconducting at a very low temperature (<10K) and cooling is therefore required. Therefore the superconducting coil is inside a cryostat filled with liquid helium with a temperature at 4.2 K. The magnetic field intensity of a superconducting magnet with Nb,Ti coils can reach up to about 15T but is clinically used up to 3T.

A more expensive material for the coils is niobium-tin (Ni3Sn) that reaches superconductivity at 18 K. This superconductor can reach magnetic field intensities of up to 25–30T at 4.2 K. It is more difficult to make the coil filament of this material, hence a combination of both materials is used and the Ni3Sn is only used in a small part of the magnet.

The magnetic field of the superconductor is always on and the only controlled way to remove the magnetic field is to slowly ramp down the power in the superconducting coils, which is a time-consuming process. The power consumption of the magnet is negligible in the steady field state. The magnetic field can also be shut down by allowing some part of the coil to become resistive. This more rapid procedure is called quenching and can be used in emergencies. It can, although rarely, also occur accidentally, for example during refilling of the cryostat. Quenching results in heat development and boil-off of the helium, hence the cryostat in superconducting magnets is always connected to outer air using a so-called quench pipe. Furthermore, a quench may be associated with significant damage to the magnet.

**Related Articles:** Electro-magnet, Magnet, Permanent magnet, Quenching, Resistive magnet, Superconductivity

**Superconductivity** *(Magnetic Resonance)* Superconductivity occurs in certain materials at extremely low temperature. The effect is characterised by close to zero electrical resistance and exclusion of the interior magnetic field.

Superconductivity occurs in some simple elements as tin or aluminium, in various metallic alloys and in heavily doped semiconductors. In a conventional superconductor (type-1 superconductor) the electrons in the electron fluid form pairs, known as Cooper pairs. The Cooper pair possesses an energy gap, ΔE, that is the minimum amount of energy that has to be applied to excite the Cooper pair. If the energy gap is larger than the thermal energy of the lattice the electrons will move through the lattice without scattering.

In the type-2 superconductor (including all known high-temperature superconductors) a small resistance, negligible in comparison with an ordinary conductor, is created by vortices in the electron fluid. When the vortices move they use some of the energy from the electron fluid and that creates the resistance. If the temperature is lowered the vortices becomes stationary and the resistance disappears.

The superconductivity appears when the temperature is below the critical temperature, Tc. In an ordinary (type-1) superconductor the critical temperature can be explained by the energy gap ΔE, see above, but in the type-2 superconductor there is no explanation (yet) to why the critical temperature appears. In Figure S.105 the resistivity, ρ, and the specific heat at constant volume, c_v, is plotted as a function of the temperature normalised to the critical temperature.

**Related Article:** Superconductive magnet
Diagnostic Radiology

Superorthicon appeared as predecessor of the image intensifier. The improved TV camera tube Superorthicon was able to directly convert this low light into video signal. This camera was used in the first x-ray fluoroscopic systems where the image was observed on a TV monitor, and not directly from the phosphorescence panel. The x-ray fluoroscopic system with Superorthicon was the first improvement of the systems with direct phosphorescence screen/panel. This way the x-ray fluoroscopic system with Superorthicon appeared as predecessor of the image intensifier.

The Orthicon uses photosensitive plate (photocathode) which emits photoelectrons and a system of electrodes to accelerate these towards the target. The scanning beam read the image directly from the target. The result is a high-resolution image. This camera has a logarithmic characteristic curve, what is well accepted by the human eye. These cameras have not been used after the 1960s, being replaced by the newer TV camera tubes Vidicon and Plumbicon (used in the x-ray fluoroscopy systems with Image Intensifier).

Related Articles: Video camera tube, Vidicon, Plumbicon


Superparamagnetic iron oxide (Magnetic Resonance) Superparamagnetism is a phenomenon in which sufficiently small particles of magnetic material behave as a single magnetic domain. In an aggregate of such particles, random orientation of the magnetic moments of the single domain particles leads to cancelling out of magnetisation in the aggregate as a whole, but in the presence of an applied magnetic field the individual moments align to generate a large magnetisation.

In MRI, particles of superparamagnetic iron oxide (usually iron (II, III) oxide, Fe₂O₃, or magnetite) are used as negative contrast agents. Related Articles: Negative contrast media, Ultrasmall particles of iron oxide (USPIO), Superparamagnetic particles

Superparamagnetic particles (Magnetic Resonance) Small particles of superparamagnetic material, normally iron oxide, are a common form of negative contrast agent in MRI.

Agents in this category include the following:

- Large superparamagnetic iron oxide preparations (SPIOs), which consist of polycrystalline particles of varying size. Administered intravenously, they have a short half life in blood and undergo phagocytosis in the liver and spleen.
- Ultrasmall superparamagnetic iron oxide preparations (USPIOs) and monocrystalline iron oxide preparations (MIOMs), consisting of smaller (<30nm) particles. They leave the vascular system slowly, often taken up by macrophages in the lymph nodes, spleen and bone marrow.
- Receptor-directed iron oxides, which are iron cores bound to receptor-specific carbohydrates.
- Antibody-labelled iron oxides, usually MION derivatives attached to antibodies or antibody fragments. An example is MION-antimyosin, which is taken up specifically by infarcted myocardium.
- Bowel suppression agents to eliminate signal from bowel which may otherwise obscure diagnosis, particularly in the presence of peristalsis.

Superior (cephalic) (General) Directional anatomical terms describe the relationship of structures relative to other structures or locations in the body. 'Superior' or 'cephalic' means above, over or towards the head (e.g. the elbow is superior to the hand).

See Anatomical relationships

Superorthicon (Diagnostic Radiology) One of the first sensitive TV camera tubes was Orthicon (successor of the Iconoscope and Emitron). This tube was more sensitive but still not suitable for the low light intensity of the fluoroscopic phosphor screen (before the use of image intensifier). The improved TV camera tube Superorthicon was able to directly convert this low light into video signal. This camera was used in the first x-ray fluoroscopic systems where the image was observed on a TV monitor, and not directly from the phosphorescence panel. The x-ray fluoroscopic system with Superorthicon was the first improvement of the systems with direct phosphorescence screen/panel. This way the x-ray fluoroscopic system with Superorthicon appeared as predecessor of the image intensifier.

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- Antibody-labelled iron oxides, usually MION derivatives attached to antibodies or antibody fragments. An example is MION-antimyosin, which is taken up specifically by infarcted myocardium.
- Bowel suppression agents to eliminate signal from bowel which may otherwise obscure diagnosis, particularly in the presence of peristalsis.
These agents produce negative contrast by shortening the $T_2$ or $T_1^*$ of hydrogen nuclei (protons) in tissue water. As the particles become larger, it is a matter of semantics as to whether the effect is regarded as a shortening of the relaxation time or irreversible dephasing due to an increase in local field inhomogeneity – i.e. a susceptibility effect.

Smaller superparamagnetic particles have a prominent $T_1$ effect as well, although this is rarely exploited in MRI.

**Related Articles:** Negative contrast media, Ultrasound particles of iron oxide (USPIO), Superparamagnetic iron oxide

### Supervised area

*(Radiation Protection)*

**Classification of Areas:** There are two types of areas where ionizing radiation may be used: controlled and supervised. The responsibility of designating an area as controlled or supervised is attributed to the registrants and licensees, who shall appoint a qualified expert, usually the radiation protection officer, to deal with the practical work. He/she shall designate as supervised any area (not already designated as controlled area) where occupational exposure conditions need to be reviewed, even though specific protection measures are not normally needed.

In determining the boundaries of any supervised area, the qualified expert (appointed by the registrants and licensees) shall take into account the magnitude of the expected normal exposures, the likelihood and magnitude of potential exposures and the nature and extent of the required protection and safety procedures. As an example, in a radiology facility, all x-ray rooms shall be controlled area while internal corridors may be a supervised area. Rooms where mobile x-ray equipment is used may also be considered a supervised area.

**Operational Conditions:** As the working conditions might change, it should be determined whether an area will be maintained as supervised or public area; the evaluation being made on the basis of regular, routine, safety assessment (including the planned use of each area) and the evaluation of shielding.

Registrants and licensees are also responsible for ensuring that: the supervised area be delineated by physical (or other suitable) means; and that warning signs and warning lights symbols (such as those recommended by International Organization for Standardization) be displayed.


### Supine

*(General)* There are a series of terms used to describe the position of an individual when undertaking different imaging examination. **Supine:** Lying on the back. For example, the position for supine abdominal imaging.

**See Patient position**

### Suppressing filter

*(General)* A suppressing filter is an electric component designed to reduce electronic transients which may be generated within electronic devices or may be carried via the mains power lines into a device.

**Contact suppressor:** When an inductive load is switched OFF it produces a large voltage spike which can be damaging. To suppress these spikes, ‘snubbers’ or contact suppressors are used, which usually are made up of a single unit containing both capacitor and a series resistor (e.g. 100 Ω and 0.1 μF) which may be connected across the switching device or load.

**Motor suppressor:** Motors are inductive and may have commutators (effectively switches), and so are a source of voltage spikes which often contain considerable energy at radio frequencies – likely to cause interference in nearby circuitry. Motor suppression may also be performed by use of a suitable RC filter combination.

**EMI filters:** Modern electronic circuitry is inherently susceptible to sharp electrical transients and electro-magnetic interference (EMI). Recent legislation requires all new devices to be capable of proper operation when subject to a certain amount of external interference, and also to operate without generating such interference themselves. It is therefore common to have an ‘EMI filter’ built into the power input circuitry to reduce incoming electrical interference, and also prevent any backward flow of interference into the mains supply. This is usually a module made of inductors and capacitors designed to block and absorb such signals.

**Surge suppressors:** where very short and very high voltage spikes might be expected, surge suppressors, usually semiconductor materials with programmed ‘breakdown voltages’, are fitted to prevent thin parallel with any load. These devices react extremely quickly and short out any overvoltage spike. Examples include varistors (MOVs), discharge tubes, and specialised diodes.

**Related Articles:** Inductor, capacitor

### Surface coil

*(Magnetic Resonance)* The term surface coil refers to a type of receive coil used in magnetic resonance imaging. These coils may consist of one or several loop shaped elements and are applied close to the surface of the part of the body that is to be examined. Contrary to a so called volume coil, such as the standard body coil in the MR-scanner, a surface coil has a more limited region of sensitivity. It ‘sees’ only the volume that is closest to each of the elements. The sensitivity is also highly dependent on the distance from the coil element and decreases as the distance increases.

Surface coils are typically more flexible than volume coils and can be applied close to the body, for example around joints, and thereby provide much better signal-to-noise ratio (SNR) than a volume coil would do. The limited region of sensitivity also contributes to the increased SNR since the coil covers a smaller tissue volume from which it can pick up thermal noise.

Many surface coils can be combined to cover large parts of the body and, depending on the number of receiver channels on the MR-scanner, they can even be used to cover the entire body.

**Related Article:** Magnetic resonance imaging (MRI)

### Surface contours

*(Magnetic Resonance)* Surface contour plots are used to represent distributions of scalar data. They can be created, for example from 3D matrices, in which there is information both about intensity (1D) and position (2D). An example is three-dimensional datasets describing temperature distributions in space, or air pressure on different locations, in which information exist both about the value of a parameter and its spatial distribution.

Surface contours are created by assigning matrix elements with identical values a specific colour while other matrix elements are transparent, thus creating a surface of points in which the values are identical (iso-surfaces). These can be overlaid as contour lines onto 2D images (like on maps) or contour surfaces onto 3D images.
Surface dose

(Radiotherapy) When exposed to ionising radiation the skin is prone to damage and an important parameter quoted in relation to this is surface dose which is commonly regarded as the dose to the surface of the skin.

Megavoltage external beam therapy with high-energy x-rays sets in motion electrons which interact with tissue over a range of several millimetres. As a result of this a dose maximum, $d_{max}$, is produced at some depth around 1–2 cm or more below the skin surface depending on the energy of the radiation beam.

To understand the significance of surface dose it is important to know the processes involved in the interaction of radiation with the structures that lie between the point of the dose maximum and the skin surface.

Early radiation damage can manifest itself in the basal layers where the dose gradient is steep. These layers lie at about 0.07 mm below the skin surface. Because of the steep dose gradient the surface dose does not give a reliable estimate of the skin dose. Late radiation damage is thought to arise from the dermis which lies about 0.05–3 mm below the skin surface where blood vessels lie which are sensitive to damage. Therefore knowledge of dose from the surface down to a depth of 3 mm is important in clinical situations.

Measurement of surface dose requires detectors which are capable of high special resolution. Metal oxide semiconductors field effect transistors (MOS-FETs) can be used which are physically small in size. Carbon loaded thermoluminescent dosimeters (TLDs) can also be used which have an effective measurement depth of 0.07 mm which corresponds to the depth of the basal layers.

In practice the several things can contribute to surface dose one of these could be the use of immobilisation devices which effectively add a layer of tissues equivalent material and push $d_{max}$ towards the skin surface. Other contributions come from electrons generated in the treatment machine. These are typically generated in the flattening filter, ionisation chamber, jaws, blocking tray and in the air between the machine head and the patient. Surface dose also increases with increasing field size. Typical values are shown in Table S.5.

**Related Article:** Build up


Surface wave

(Ultrasound) When an ultrasound wave propagates in a medium, the particles within the medium will start oscillating. In gases, liquids and soft tissue this oscillation is always in the same direction as the ultrasound wave, giving rise to longitudinal wave propagation. However, in solid materials both longitudinal (compressional) and transverse (shear) wave propagation are possible. In a transverse wave the particles oscillate perpendicular to the wave propagation direction. Surface (Rayleigh) waves are a combination of longitudinal and transverse waves and can propagate at the boundary between a liquid and a solid, Figure S.106. Ocean waves are surface waves.

**Related Articles:** Longitudinal wave, Transverse wave, Lamb wave

Survey meter

(Radiation Protection) A calibrated radiation detection instrument used to make accurate measurements of radiation fields in order to make dose assessments of the output of clinical equipment, or to describe the hazard to persons present in the area.

Surviving fraction

(Radiotherapy) The surviving fraction is the proportion of cells that survive irradiation. A plot of surviving fraction against radiation dose is called a cell survival curve and is usually presented in the form shown in Figure S.107 for x- and γ-rays, with dose plotted on a linear scale and surviving fraction on a logarithmic scale.

A cell’s radiosensitivity can be determined from cell survival curves. The shape of survival curves is commonly described by the linear-quadratic model with the ratio of its two parameters, α and β, often used as a measure of a cell’s radiosensitivity. An alternative parameter of radiosensitivity relevant to the doses used in fractionated radiotherapy is the surviving fraction at the 2 Gy level (SF₂). This has the advantage of only requiring measurement at one point on the survival curve, useful for human tumours/tissues where samples are often small and difficult to obtain.

**Abbreviation:** SF₂ = Surviving fraction at 2 Gy level.

---

**TABLE S.5**

<table>
<thead>
<tr>
<th>Energy</th>
<th>Surface Dose % of $d_{max}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.25 MeV γ-rays</td>
<td>23</td>
</tr>
<tr>
<td>4 MV</td>
<td>18</td>
</tr>
<tr>
<td>6 MV</td>
<td>15</td>
</tr>
</tbody>
</table>

**FIGURE S.106** Surface (or Rayleigh) wave. The particles move in a circular pattern. (Graphs courtesy of EMIT project, www.emerald2.eu)

**FIGURE S.107** Shape of survival curve for mammalian cells exposed to radiation. The fraction of cells surviving is plotted on a logarithmic scale against the radiation dose on a linear scale.
**Susceptibility**

*(Magnetic Resonance)* The degree of magnetisation of a material in an applied magnetic field is called magnetic susceptibility, \( \chi \). The magnetic field is strengthened, \( \chi \) positive, by a paramagnetic material but it is weakened, \( \chi \) negative, by a diamagnetic material.

Susceptibility effect can be advantageous in the following cases:

1. Detection of haemorrhage due to the paramagnetic properties of deoxyhaemoglobin.
2. Dynamic susceptibility contrast MRI, where measurement of perfusion can be assessed using a paramagnetic contrast agent (gadolinium chelate). The contrast agent is injected intravenously and followed through rapid imaging. The dynamic time curve describing susceptibility-induced signal drop during first passage of the contrast agent can then be used to calculate perfusion parameters such as cerebral blood flow (CBF) using tracer kinetics.

**Related Articles:** Dynamic susceptibility contrast MRI, Ferromagnetism, Paramagnetism

**Swept gain**

*(Ultrasound)* Swept gain control, sometimes called 'depth-corrected gain control', 'time-corrected gain control' or 'distance-amplitude-correction gain control', is a scheme designed for increasing the gain of the receiver system as the later the ultrasound echoes are received. By adjusting this gain properly, echoes from acoustic mismatches of the same magnitude but at different depths will produce spikes of identical amplitude on the cathode ray tube screen irrespective of depth of the interface that produced the echo.

**Related Articles:** Depth gain control, Time gain control

**Switch off**

*(General)* See *Switch on*

**Switch on**

*(General)* Switching on means turning on (energizing) an electrical device (or vice versa—off) by allowing current to flow through the circuit and the switch when closed (or by providing isolation when opened).

In IEC-60601-1 compliant medical electrical equipment the switches used to control power to medical electrical equipment or its parts, including mains switches, have their 'on' and 'off' positions marked with symbols correspondingly by IEC 60417-5007 (I) and IEC 60417-5008 (O), or indicated by an adjacent indicator light, or indicated by other unambiguous means (Figure S.108).

**Related Articles:** Switch off, Switch, Toggle switch, Dead man's switch, Exposure switch, Foot switch

**Symmetric energy window**

*(Nuclear Medicine)* In a symmetric energy window, the photopeak is located in the centre of the window. For example, \(^{99m}\)Tc has a photopeak at 140 keV. A symmetric energy window with 20% would result in a 140 ± 14 keV window. Changing the width of the energy window to a narrower window will degenerate the detector uniformity. In SPECT and planar scintillation camera imaging an appropriate energy width is decided with a uniformity quality control provided by the manufacturer. A correct and stable energy window is crucial for accurate data acquisition.


**Hyperlink:** [www.IAEA.org](http://www.IAEA.org)

**Synchronisation**

*(General)* Synchronisation is a situation when two or more processes coordinate their activities based upon a condition. Synchronisation is also the process of determining (usually channel) parameters from a received signal (e.g. carrier frequency offset, carrier phase, or symbol timing), or timekeeping, which requires the coordination of events to operate a system in unison (based on GPS and network time protocol (NTP) timekeeping systems).

In diagnostic radiology, reduction of breathing motion artefacts can be achieved by using respiration synchronised gating techniques. The latter use external devices to predict the phase of the respiration cycle while the patient breaths freely.

Cardiac synchronisation methods (see the article *Cardiac gating*) used to synchronise with cardiac motion essentially rely on the electrocardiogram (ECG). The peripheral pulse is only used as a last resort. Cardiac synchronisation limits the artefacts linked to the motion of the heart and blood flow, thus enabling the different phases of the cardiac cycle to be sampled.

**Related Article:** Cardiac gating

**Syringe shield**

*(Nuclear Medicine)* Syringe shields are used in nuclear medicine and nuclear pharmacy to reduce the absorbed dose to fingers and hands of the persons working with both preparation and patient injections of the radiopharmaceuticals. One type of syringe shield is a tungsten shield with a leaded-glass viewing window for reading syringe markings. Another syringe shield is made entirely of leaded glass facilitating the reading of the syringe marks. A thickness of
3 mm lead reduces the $^{99m}$Tc-exposure rate by a factor of 1000. Tungsten is 50% denser than lead thus reducing the syringe shield thickness. The relative absorbed dose to the fingers is also dependent on the position on the shield and on the filling volume of the syringe (Figures S.109 and S.110).


**System of work**

(Radiation Protection) A ‘system of work’ is a term used in certain countries to describe radiation protection procedures regarding access to designated radiation areas, to minimise the risk of staff and other persons receiving significant radiation exposure.

Systems of work contain detailed information and instructions for any worker not designated as a classified worker, and for contractors and visitors who are required to enter a controlled area, such that their radiation doses received are kept below 3/10 of any dose limit.

In principle classified workers, who are by definition expected to receive more than 3/10 of a dose limit, cannot be following the system of work for the controlled area because if they did then they would not have to be classified. They will still be expected to follow the more general requirements set out in the local rules document.

Information on any safety requirements, time restrictions, use of personal protective equipment and devices, shielding etc. should be specified in the systems of work. Usually the local radiation protection supervisor, with support from the head of department and radiation protection adviser is given the duty of preparing the system of work, which will be tailored to various functions and situations. It is a further requirement that all involved individuals must receive adequate training in the procedures set out in the document.

Finally, there may be contingencies described in the local rules document for the controlled area that are used for emergency situations (fire alarm, etc.), and for which in the interests of the immediate danger to staff and other persons, the normal systems of work will not therefore apply.

**Related Articles:** Classified worker, Controlled area, Local rules, Radiation protection supervisor, Radiation protection adviser

**System resolution in a scintillation camera**

(Nuclear Medicine) A gamma camera’s ability to depict sharp edges and point sources is referred to as the system resolution. There are two main factors affecting the system resolution $R_{sys}$ in a gamma camera, namely the intrinsic resolution $R_{int}$ and the collimator resolution $R_{coll}$. The intrinsic resolution includes the spatial degenerative properties of the electronics and detector crystal. The combination of these two factors is

$$R_{sys} = \sqrt{R_{int}^2 + R_{coll}^2} \quad (S.21)$$

Since the collimator resolution depends on the source to collimator distance so does the system resolution. At typical organ depths (5–10 cm) the system resolution is much poorer than the intrinsic resolution which means that it is primarily determined by the
collimator resolution. Differences in intrinsic resolution between camera systems might appear when imaging superficial structures but is not a dominating factor when imaging organs and structures at greater depths.

The system resolution is also degraded by scattered radiation and septal penetration. The fraction of scattered radiation increases with photon energy and distance travelled through an absorbing medium before detection. The effects of scattered radiation can be limited using pulse height analysis and dual energy window techniques so that the degeneration of system resolution is small.

The fraction of septal penetration increases with photon energy. This contribution to resolution degeneration can be limited using a collimator appropriately designed for the specific photon energy.

**Related Articles:** SPECT, Spatial resolution, Spatial resolution


**Système International (SI)**

(Radiation Protection) The International Bureau of Weights and Measures (Bureau International des Poids et Mesures–BIPM) during several General Conferences on Weights and Measures (Conference Générale des Poids et Mesures–CGPM) adopted the International System of Units (SI) (Système International) based on the seven units: metre (m) for length, kilogram (kg) for mass, second (s) for time, ampere (A) for electric current, kelvin (K) for thermodynamic temperature, candela (cd) for luminous intensity and mole (mol) for amount of a substance, and on two supplementary units: radian (rad) for plane angle and steradian (sr) for solid angle.

The SI introduced the names of multiples and submultiples of the units which are formed by means of the following prefixes:

<table>
<thead>
<tr>
<th>Prefix</th>
<th>Symbol</th>
<th>Exponent</th>
</tr>
</thead>
<tbody>
<tr>
<td>yotta</td>
<td>Y</td>
<td>10^24</td>
</tr>
<tr>
<td>zetta</td>
<td>Z</td>
<td>10^21</td>
</tr>
<tr>
<td>exa</td>
<td>E</td>
<td>10^18</td>
</tr>
<tr>
<td>peta</td>
<td>P</td>
<td>10^15</td>
</tr>
<tr>
<td>tera</td>
<td>T</td>
<td>10^12</td>
</tr>
<tr>
<td>giga</td>
<td>G</td>
<td>10^9</td>
</tr>
<tr>
<td>mega</td>
<td>M</td>
<td>10^6</td>
</tr>
<tr>
<td>kilo</td>
<td>k</td>
<td>10^3</td>
</tr>
<tr>
<td>hecto</td>
<td>h</td>
<td>10^2</td>
</tr>
<tr>
<td>deca</td>
<td>da</td>
<td>10^1</td>
</tr>
<tr>
<td>yocto</td>
<td>y</td>
<td>10^-24</td>
</tr>
<tr>
<td>zepto</td>
<td>z</td>
<td>10^-21</td>
</tr>
<tr>
<td>atto</td>
<td>a</td>
<td>10^-18</td>
</tr>
<tr>
<td>femto</td>
<td>f</td>
<td>10^-15</td>
</tr>
<tr>
<td>pico</td>
<td>p</td>
<td>10^-12</td>
</tr>
<tr>
<td>nano</td>
<td>n</td>
<td>10^-9</td>
</tr>
<tr>
<td>micro</td>
<td>μ</td>
<td>10^-6</td>
</tr>
<tr>
<td>milli</td>
<td>m</td>
<td>10^-3</td>
</tr>
<tr>
<td>centi</td>
<td>c</td>
<td>10^-2</td>
</tr>
<tr>
<td>deci</td>
<td>d</td>
<td>10^-1</td>
</tr>
</tbody>
</table>

The SI consists of the coherent-derived units expressed in terms of base units, e.g. velocity (m s^-1), acceleration (m s^-2) and coherent derived units with special names, e.g. hertz (Hz) for frequency, newton (N) for force, joule (J) for energy, becquerel (Bq) for activity, gray (Gy) for absorbed dose, sievert (Sv) for equivalent dose, effective dose, ambient dose equivalent and directional dose equivalent.

**Abbreviations:** BIPM = Bureau International des Poids et Mesures and SI = International System of Units (Système International).


$T_1$ (Magnetic Resonance) See Relaxation time

$T_1$-weighted (Magnetic Resonance) In a $T_1$-weighted MR image the difference in $T_1$-relaxation time is the main parameter of the image contrast. However the proton density (PD) will always contribute to the image intensity and thus also influence image contrast.

In spin echo MRI $T_1$-weighting can be achieved by using a short repetition time (TR) and a short echo time (TE), typically TR < 700 ms and TE < 30 ms. With the short TR the spins in the tissues with a long $T_1$ will not have time to fully relax back to the longitudinal direction ($z$-direction). This leads to a reduction in signal for tissues with long $T_1$ because there are not so many spins in the longitudinal direction that can be excited. The short TE will reduce the influence of $T_1$ relaxation.

In spin echo MRI the transversal magnetisation can be described by

$$M_{xy} = M_0 (1 - e^{-TR/T_1}) e^{-TE/T_2}$$

If TE is much shorter than $T_2$, the equation can be reduced to

$$M_{xy} = M_0 (1 - e^{-TR/T_1})$$

For small TR the equation can be reduced further (ignoring the higher order terms of a Taylor series representation of the exponential term):

$$M_{xy} = M_0 \frac{TR}{T_1}$$

The last equation shows that the transversal magnetisation (signal) depends on the $T_1$ and the spin density ($M_0$) of the tissue.

In the $T_1$-weighted image fat has a strong signal due to its short $T_1$ constant, since a larger percentage of fat spins will relax before the next excitation RF pulse. Water has a rather long $T_1$ constant and will therefore have more signal left and appear white in the image. Cerebral white matter has a shorter $T_1$ than grey matter and will therefore be more intense (Figure T.1).

In spoiled gradient echo MRI strong $T_1$-weighting can be achieved by using a large flip angle (e.g. >50°) to avoid the possibility for the excited spins to have time to return to the longitudinal direction.

Related Articles: Echo time (TE), Relaxation, Relaxation time ($T_1$), Repetition time (TR), $T_1$


$T_2$ (Magnetic Resonance) $T_2$ (or $T_2^*$) denotes the transverse relaxation time, that is the time constant for the irreversible decay of the magnetisation vector $M$ component that is perpendicular to the main magnetic field $B_0$ after RF excitation. See Relaxation time.

Related Articles: Relaxation, Relaxation time, Relaxation rate, $T_2$-weighted

$T_2$-shine through (Magnetic Resonance) The $T_2$-shine through effect is especially important when diagnosis is based on diffusion weighted (DW) images, that is if ADC maps not are calculated. In diffusion imaging, the basic protocol uses spin echo (SE) pulse sequence with long echo times (TE) and long repetition times (TR), a combination that normally constitutes a $T_2$-weighted image. On top of this, the diffusion encoding gradients are added, (requiring substantially long TE's) and thus the signal decay is influenced by the combined effects of $T_2$-relaxation and long TE's as well as the diffusion coefficient ($D$) and the diffusion sensitivity ($b$-value). The signal decay is given by

$$S = S_0 e^{-T_2/TE} e^{-bD}$$

where

- $S_0$ is the initial signal amplitude
- $S$ is the measured signal affected by the imaging protocol parameters TE and $b$-value as well as the intrinsic object parameters $T_2$ and $D$

The equation implies that for low diffusion sensitivities the $T_2$ effect will dominate. The actual crossing point where the $T_2$-weighted contrast will no longer dominate the image depends on the $T_2$ and $D$ values of the tissue. An intense signal area can for an image acquired with a low $b$-value indicate either a long $T_2$ or decreased diffusion. Examining for instance a patient with an ischemic stroke (Figure T.2) in the acute phase and who has a history of earlier strokes, lesions corresponding to the ‘old infarct’ will also appear bright in images of low diffusion sensitivity due to its high water content. In the upper row a high signal lesion is marked with an arrow in the $b = 0$ s/mm$^2$ mage. A similar-looking lesion marked with an arrow, is also present in the bottom row. Following the lesions for increasing $b$-values shows that for a $b$-value of 1000 s/mm$^2$ the lesion corresponding to a chronic lesion, (bottom row) is no longer visible. Examining the ADC maps gives a correct representation of the lesions where the acute lesion show a low ADC value whereas the chronic lesion appears bright due to increased water content.

For ischaemic stroke it is considered safe to use $b$-values above 1000 s/mm$^2$ in order to avoid effects from $T_2$-shine through (Figure T.2).

Related Articles: $b$-value, Diffusion imaging

$T_2$-weighted (Magnetic Resonance) In a $T_2$-weighted MR image differences in $T_2$-relaxation times are the main source of image contrast. However the proton density (PD) will always contribute to the image intensity and thus strongly influence image contrast.

In a conventional spin echo sequence this can be achieved with a long repetition time (TR > 1500 ms) and an appropriate long echo time (TE > 75 ms). With a long TR the excited spins in all tissues will have time to relax back to the longitudinal direction ($z$-direction). Therefore, image contrast is based on differences in $T_2$-relaxation and spin density $M_0$, see Figure T.1.
In spin echo MRI, the transversal magnetisation (excited spins) can be described by
\[ M_{xy}^{te} = M_0 \left(1 - e^{-TR/T_1}\right) e^{-TE/T_2} \]

If TR is much longer than \( T_1 \) the equation is reduced to
\[ M_{xy}^{te} = M_0 e^{-TE/T_2} \]

The signal in the \( xy \)-direction is only dependent on the spin density \( M_0 \) and the \( T_2 \) of the tissue.

In the \( T_2 \)-weighted image fat, that has a short \( T_2 \) constant so that signal \( (M_{xy}) \) drops off faster after the RF-excitation, has a weak signal on \( T_2 \)-weighted image with longer TE and will appear with less signal intensity. Water has a rather long \( T_2 \) constant and will therefore have much signal left and appear white in the image. Cerebral white matter is less intensive than grey matter.

In a spoiled gradient echo there is no refocusing of the spins (see Spin echo), which leads to the effect that inhomogeneities in the magnetic field will reduce the transversal relaxation rate. Hence, true \( T_2 \)-weighting is not achieved in this type of sequences but rather so-called \( T_2^* \)-weighting (see Relaxation).

**Related Articles:** Echo time (TE), Flip angle, Relaxation, Relaxation time, Spin echo, Repetition time (TR), \( T_2 \)


\( T_2^* \) (Magnetic Resonance) Transverse magnetisation generated by a RF pulse undergoes exponential decay because of spin–spin relaxation. However, in practice it is found that the free induction decay (FID) or echo signal decays more rapidly than \( T_2 \) alone would suggest. The actual rate of decay is known as \( T_2^* \), such that
\[ M_{xy} = M_0 \exp(-t / T_2^*) \]

This phenomenon occurs because of inhomogeneities on a microscopic scale in the static magnetic field. Even if a magnet could be constructed with a perfectly uniform field, this would be distorted by placing an object in the field for imaging, and specifically by differences in magnetic susceptibility between components of the object (different tissues and anatomical structures...
in the case of in vivo imaging). The resulting effective relaxation time is given by

$$\frac{1}{T_2} = \frac{1}{T_2} + \frac{\Delta \omega}{2}$$

where $\Delta \omega = \gamma \Delta B$ and $\Delta B$ is the magnetic field inhomogeneity.

A spin echo can be used to reverse the effects of field inhomogeneity and recover a true $T_2$-weighted signal. Alternatively, a sequence using gradient echo acquisition alone will result in a $T_2^*$-weighted image.

**Related Articles:** Free induction decay (FID), Gradient echo (GE), Magnetic susceptibility, Spin echo, Spin–spin relaxation, Static field, Transversal magnetisation, $T_2$-weighted

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**Table top**

*(Diagnostic Radiology)* The top of the patient table of any imaging equipment has to be not only capable to support the patient, but also to have minimal influence on the image, that is be almost ‘invisible’. For example, the patient table in x-ray imaging has to be with minimal absorption, as it stays between the patient and the detector (usually the cassette holder and anti-scatter grid are under the table top). In contemporary x-ray equipment this is achieved by using table top made of special carbon fibre materials. However even such a table top would absorb 30%–50% of the radiation exiting the patient (the modulated x-ray beam), thus reducing significantly the amount of modulated radiation reaching the detector. To reduce this effect (as well as the magnification), some x-ray radiography procedures require the detector (or film) to be as close as possible to the patient (e.g. typically in mammography). In this case the detector (or film cassette) is placed below the patient – directly on the top of the table. This radiographic method is known as ‘table top examination’.

**Tangent fields**

*(Radiotherapy)* This is the most common field arrangement used for breast radiotherapy. An illustration of this is given in Figure T.3. In this technique the diverging posterior beam edges of two fields are matched together along a single line to minimise the dose to the lung tissue, by careful selection of gantry angle.

**Related Articles:** Beam arrangement, Beam divergence

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**Tantalum**

*(General)*

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Ta</th>
</tr>
</thead>
<tbody>
<tr>
<td>Element category</td>
<td>Transition metal</td>
</tr>
<tr>
<td>Mass number $A$</td>
<td>181</td>
</tr>
<tr>
<td>Atomic number $Z$</td>
<td>73</td>
</tr>
<tr>
<td>Atomic weight</td>
<td>180.948 g/mol</td>
</tr>
<tr>
<td>Electronic configuration</td>
<td>$1s^2 2s^2 2p^6 3s^2 3p^6 3d^{10} 4s^2 4p^6 4d^{10}$ $4f^{14} 5s^2 5p^6 5d^{10} 6s^2$</td>
</tr>
<tr>
<td>Melting point</td>
<td>3290 K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>5731 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>16.4 g/cm$^3$</td>
</tr>
</tbody>
</table>

**History:** Tantalum was first discovered in 1802, but it was only in 1907 that a sample of the relatively pure metal was produced. It is usually extracted along with niobium from minerals such as columbite and tantalite. It is a strong ductile metal with exceptional resistance to chemical attack. It is therefore used to make components that need to operate in aggressive environments where corrosion is not acceptable.

**Medical Applications:** Instruments and implants – The passive nature and corrosion resistance of tantalum make it a suitable material for surgical instruments and implants. In particular it is regarded as a favourable material for implants in bone, where it has been reported to display osseointegration.

**TAR (tissue air ratio)** *(Radiotherapy)* See Tissue air ratio (TAR)

**Target angle** *(Diagnostic Radiology)* See Anode angle

**Target cooling** *(Diagnostic Radiology)* See Anode-cooling curve

**Target dose** *(Radiotherapy)* This could refer to many different aspects such as mean target dose, maximum target dose, minimum target dose, modal target dose. The target dose would be considered along with dose–volume constraints.

**Related Articles:** Mean target absorbed dose, Maximum target absorbed dose, Minimum target absorbed dose, Modal target absorbed dose

**Target dose distribution** *(Radiotherapy)* The distribution of dose within a target is called the target dose distribution. Knowledge of this allows the degree of conformity and the homogeneity of the dose distribution pattern inside the radiotherapy target volume to be evaluated.

**Target film distance (TFD)** *(Diagnostic Radiology)* The term target film distance (TFD) is used in planar x-ray radiology to describe the distance between the x-ray target or focal spot in the x-ray tube, and the surface of the film. It is sometimes referred to as the focal film distance (FFD). In modern digital radiology where film is no longer used, the distance between the target and detector is called the focal detector distance (FDD) which is also known as the target detector distance (TDD), the focal image receptor distance (FID), or the target image receptor distance (TID) *(Figure T.4).*

**Related Articles:** Focal film distance (FFD), Object film distance (OFD), Object image receptor distance (OID), Magnification

**Target localisation** *(Radiotherapy)* The first stage in planning radiotherapy treatment is to determine the location of the tumour/target. This is generally...
achieved using imaging of soft tissue to establish the location and visible extent of the tumour. This is known as the GTV (gross tumour volume). Margins are grown around this to account for uncertainties in the treatment process.

**Localisation for Planning:** CT and MRI are the imaging techniques used most commonly for target localisation in radiotherapy planning. The location of the tumour and critical normal tissues is outlined on the images and dose calculations carried out monitoring the doses each are expected to receive.

**Localisation for Verification:** Soft tissue imaging may be used during each treatment fraction to locate the tumour position and correct set-up before irradiation starts. A common technique to achieve this is conebeam CT on the treatment machine.

See also all other articles on volumes and margins

**Abbreviation:** GTV = Gross tumour volume.

**Hyperlinks:** [http://www.icru.org/](http://www.icru.org/) ICRU website

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**Target organ**

(Radiotherapy) Radiotherapy treatments may target a particular organ; however, it is more common for them to be described in terms of target volumes – for example the clinical target volume (CTV) describes the full extent of malignant growth that is grossly visible, plus subclinical microscopic growth which is based not just on anatomy but biological considerations. The CTV often includes regional lymph nodes.

**Related Articles:** Target volume, Clinical target volume

**Target volume**

(Radiotherapy) This is a generic term for the volume containing malignancy that is targeted by radiotherapy or brachytherapy in order to eliminate the cancerous cells. More specific descriptors are given in ICRU 60, specifying distinct components of the target volume as the gross tumour volume (GTV), clinical target volume (CTV), and planning target volume (PTV). See individual articles for more information.

**Related Articles:** Gross tumour volume (GTV), Planning target volume (PTV), Clinical target volume (CTV)


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**Target of x-ray tube**

(Diagnostic Radiology) The small region of the anode, which is bombarded by the thermal electrons and produces x-rays is called the target. In stationary anode x-ray tubes it is a small Tungsten plate (~1 mm thick). Almost 99% from the energy imparted to the target by the electrons is converted to heat and generation of secondary electrons (strictly speaking the thermal energy converted directly to heat is ~75%). Due to this reason the material of the target is normally tungsten – a material with a very high melting point (3410°C) and high atomic number. The latter is important for the effective conversion the energy of electrons to x-rays. The ‘Bremsstrahlung generation efficiency’ (η) is the ratio of the intensity of x-ray radiation (W) and the energy flux of the electron beam (E ∼ I U).

The intensity of x-ray radiation W (x-ray energy flux density) is W ∼ I U² Z, where W is the intensity of x-ray radiation, I is the
anode current, Z is the atomic number of anode, U is the accelerating high voltage (anode tension – kV), from what follows that \( \eta \sim \frac{W/E}{k U Z} \).

The constant \( k \) is experimentally established as \( 1.1 \times 10^{-9} \). For tungsten \( Z = 74 \). Thus for 100 kV the bremsstrahlung generation efficiency is approximately 0.8%. But the x-rays are spread in different directions (isotopic x-ray emission) and just a small proportion of them leave the tube in the direction of the patient. This radiation is additionally absorbed in the tube housing and added filtration. Thus less than 0.1% of the energy imparted to the anode is converted to useful radiation. Due to this reason a very high electron beam energy is necessary for production of useful x-rays. As a result the target of the anode heats up to very high temperatures during the exposure (Figure T.5 – the photograph shows the anode removed from the tube).

The other reason for choosing tungsten for the anode material is its high thermal conductivity. One disadvantage of tungsten is thermomechanical stress due to high thermal gradients, which leads to cracks on the tungsten surface. The cracks not only decrease the life of the x-ray tube (damaging the anode), but also make the target surface uneven. The cracked anode surface causes scattering or absorbing (in the cracks) of some of the x-rays quanta, and so decreases the tube efficiency. In order to minimise this effect rhenium is added to the tungsten target (approximately 2%–10% Re/W alloy) and special care is taken to remove the heat accumulated in the anode. The x-ray tubes with rotating anode have all their front surfaces covered with this alloy, although only part of it is bombarded – the thermal path (Figure T.6 – the photograph shows the anode removed from the tube).

Some x-ray tubes use an anode target made from other materials (e.g. molybdenum or rhodium, used for mammographic tubes). These types of target produce significant percentage of characteristic radiation, which forms part of the mammographic x-ray spectrum.

Related Articles: Anode, Rotating anode, Stationary anode, Filament heating

Hyperlinks: EMERALD (DR module), www.emerald2.eu.

Targeted alpha therapy
(Radiotherapy)

Background: Radioisotopes (radionuclides) are used in nuclear medicine procedures for imaging and therapy. Imaging isotopes emit gamma rays; therapeutic isotopes emit low energy gamma rays or high energy beta radiation. Cancer is the main target for therapeutic nuclear medicine. A new approach to therapy is emerging where radioisotopes that emit very short range (80 μm) alpha particles are tagged onto monoclonal antibodies for targeted alpha therapy (TAT). The alpha radiation is high linear energy transfer (LET) radiation and transfers ~100 keV/μm to the targeted cells, causing increased double strand breaks in the nuclei of the targeted cancer cells. The radiation weighting factor for alphas is 20 and the relative biological effectiveness (RBE) for tumour regression is ~3–5.

As of 2009, clinical trials are in progress for acute myelogenous leukaemia (AML), metastatic melanoma, lymphoma, glioblastoma multiforme (GBM), and breast and prostate bone metastases.

Methods: There are a number of suitable alpha emitting radioisotopes, viz. \(^{188}\)Tb, \(^{211}\)At, \(^{212}\)Bi, \(^{213}\)Bi, \(^{222}\)Ra, \(^{212}\)Ra, \(^{225}\)Ac but the most important of these would be the \(^{225}\)Ac: \(^{213}\)Bi generator. With a half life of 10 days, \(^{225}\)Ac can be delivered around the world and eluted (milked) to produce \(^{213}\)Bi. This is chelated to the monoclonal antibody specific for the targeted cancer to form the alpha-immunoconjugate (AIC). \(^{213}\)Bi has a 46 min half life, so patients are treated as outpatients.

The key requirement is to achieve disease regression within the maximum tolerance dose (MTD). This has been achieved in the majority of trials.

Comment: TAT was indicated for micrometastases or subclinical cancer, for which targeting would be swift and the short alpha range effective in killing the cancer cells. Liquid cancers such as leukaemia are also indicated as the uptake of short-lived radioisotope in the cancer mass is achieved within 5 min. However, TAT was not indicated for solid tumours, where the diffusion time for monoclonal antibodies can be 24–24 h, and only a heterogeneous distribution would be achieved. The short alpha range would prohibit effective cross fire, as is the case for beta rays. In the case of GBM, the AIC is injected into the post-surgical cavity, so shallow diffusion of the conjugate can be readily achieved.

Abbreviations: AIC = Alpha-immunoconjugate, AML = Acute myelogenous leukaemia, GBM = Glioblastoma multiforme, LET = Linear energy transfer, MTD = Maximum tolerance dose and TAT = Targeted alpha therapy.

Related Article: Alpha-immunoconjugate

Tc-99m albumin microspheres (HAM)
(Nuclear Medicine) Tc-99m-labelled human albumin microspheres (HAM) are albumin spheres normally with an average
diameter of 40 μm (range 10–50 μm). It is used clinically for scintigraphy of pulmonary perfusion and determination of right-to-left shunt after an intravenous injection. After an arterial injection microspheres may be used for regional perfusion in other organs.

A kit contains human serum albumin (HSA) microspheres, stannous chloride and surfactants to avoid aggregation. Different labelling techniques may be used. The specific activity of 99Tc-HAM should be higher than 185 MBq per 10 μm microspheres and the number of 99Tc-atoms in the 99Tc-eluete should be minimized.

The number of microspheres can be determined by Coulter counter measurements. The 99Tc-HAM microspheres should be homogenously suspended to avoid in vivo aggregates when injected intravenously. Thus, no blood aspiration into the syringe is allowed. The radiochemical purity of 99Tc-HAM microspheres can be determined using membrane (Millipore) filtration with a pore diameter of 3 μm. Less than 5% of the activity should be measured on the Millipore filter. It is also possible to measure the radiochemical purity by centrifugation, or more conveniently by paper or thin-layer chromatography.

99Tc-HAM microspheres are trapped by the capillary bed, that is the lung after an IV. injection where they later are metabolised – larger microspheres are removed more slowly than the small one.

The organs that are most exposed to radiation are the lungs (mGy/MBq) and bladder wall (mGy/MBq). The retention in the lung can be modelled using a double exponential, 1.8h (60%) and 1.5d (40%). The microspheres are primarily excreted by the kidneys. The effective dose is 11 μSv/MBq – one of the lowest in diagnostic imaging. Normally the administered activity to an adult patient is 185 MBq.

**Abbreviations:** HAM = Human albumin microspheres, HSA = Human serum albumin and IV = Intravenous injection.

**Related Articles:** Tc-99m-sodium pertechnetate, Tc-99m-HMPAO.


**Tc-99m SestaMIBI (methoxyisobutylisonitrile)**

(Tc-99m SestaMIBI, Myoview™ (GE Healthcare) is used for myocardial perfusion studies in patients with coronary artery disease to diagnose ischemic heart disease and reduced regional perfusion and to localize myocardial infarction and perfusion defects. The uptake of the lipophilic 99mTc-tetrofosmin complex in the heart muscle is proportional to the blood flow. Uptake is due to diffusion and the complex is retained by the viable myocytes.

Two injections are usually required to distinguish between a transient and a permanent perfusion defect. One examination is performed after maximum exercise or pharmacological stress and the other one at rest. The examinations are performed using SPECT or a planar technique and commence at 15–30 min after injection but the delay can be up to 4h. Stress-rest imaging can be performed using either 1- or 2-day protocols.

The MIBI kit contains lyophilised, sterile, pyrogen-free, substance in nitrogen atmosphere, prepared for labelling with 1–3 mL of 99mTc–sodium pertechnetate (0.925–5.55 GBq). The vial should be agitated vigorously to dissolve the material and then be placed in a boiling water bath for 10 min. The preparation is ready for use after being cooled at room temperature at about 15 min and should be stored at room temperature (15°C–25°C) up to 6h protected from light.

Small amounts of 99mTc – impurities may be formed during labelling, which is taken up, for example in the thyroid and the liver. The radiochemical purity shall be at least 94% and may be checked using instant thin-layer chromatography (ITLC).

The injected activity is rapidly eliminated from the circulating blood with 2.5% of injected activity 10 min after injection. Technetium-99m-MIBI is mainly accumulated in muscle tissue and the myocardial uptake is 1.0%–1.4%. The initial high liver uptake the radiopharmaceutical excreted via the hepatobiliary tract and the faeces is 37% in 24h and via the urinary tract 29%.

The highest absorbed doses are received by the gallbladder wall, the kidneys and the intestinal tract. The calculated effective dose is 0.0085 mSv/MBq at rest and 0.0075 mSv/MBq at stress.

**Related Articles:** Radionuclide generator, Tc-99m tetrofosmin.


**Tc-99m tetrofosmin**

(Nuclear Medicine) Technetium-99m-tetrofosmin, Myoview™ (GE Healthcare) is used for myocardial perfusion studies in patients with coronary artery disease to diagnose ischemic heart disease and reduced regional perfusion and to localize myocardial infarction and perfusion defects. The uptake of the lipophilic 99mTc-tetrofosmin complex in the heart muscle is proportional to the blood flow. Uptake is due to diffusion and the complex is retained by the viable myocytes.

Two injections are usually required to distinguish between a transient and a permanent perfusion defect. One examination is performed after maximum exercise or pharmacological stress and the other one at rest. The examinations are performed using SPECT or a planar technique and commence at 15–30 min but the delay can be up to 4h. Stress-rest imaging can be performed using either 1- or 2-day protocols.

The tetrofosmin kit contains lyophilised, sterile, pyrogen-free, substance in nitrogen atmosphere, prepared for labelling with 4–8 mL of 99mTc–sodium pertechnetate not exceeding 1.11 GBq/mL and 8.88 GBq. The radiopharmaceutical is ready for use after 15 min reaction in room temperature and should be used within 8h after preparation. The preparation should be stored at 2°C–8°C.

Small amounts of 99mTc impurities may be formed during labeling, which is taken up, for example in the thyroid and the liver. The radiochemical purity shall be at least 90% and may be checked using instant thin-layer chromatography (ITLC).

The injected activity is rapidly eliminated from the circulating blood with less than 5% of injected activity remaining 10 min after injection. Tetrofosmin is mainly accumulated in muscle tissue. The maximum uptake in heart muscle is approximately 1.2% slowly decreasing to about 0.7% at 4 h p.i. The excretion via the hepatobiliary tract and the faeces is 34% in 48h and via the urinary tract 39%.
The highest absorbed doses are received by the gallbladder wall, the intestinal tract, the uraery bladder wall and the kidneys. The calculated effective dose is 0.0089 mSv/MBq at rest and 0.0071 mSv/MBq at stress.

**Abbreviation:** p.i. = Post-injection.

**Related Articles:** Radionuclide generator, Tc-99m SestaMIBI (methoxyisobutyl isonitride)


**Tc-99m-albumin (HSA)**

*(Nuclear Medicine)* Technetium-99m-albumin, \(^{99m}\text{Tc-\text{HSA}}\), is used for static or gated cardiac blood pool imaging, first-pass studies, or regional circulatory imaging. The time of examination is immediately or shortly after IV administration. Recommended activity is 111–185 MBq for blood pool imaging, 370–450 MBq for angiocardiography, 185–925 MBq for gated ventriculography, and 18.5–185 MBq for blood flow studies.

Human serum albumin, HSA kits contain HSA and Sn\(^{2+}\). HSA is the most common protein in the blood and is isolated from donor blood. Several methods for labelling have been derived and commercial products available are Albumoscan (Nordion), TechneScan (Mallinkrodt/Tyco) and VesculoCis TCK-2 (CIS Bio).

The kit contains lyophilised, sterile, pyrogen-free, substance in nitrogen atmosphere, prepared for easy labelling with \(^{99m}\text{Tc}\) – sodium pertechnetate by adding 1–8 mL with an activity up to 2.22 GBq. After preparation, the injection solution should be used within 6 h. The radiochemical purity can be tested by paper or thin-layer chromatography (recommended), and the purity should not be less than 95%.

\(^{99m}\text{Tc}\)-HSA is distributed homogeneously in the vascular component following an IV injection. It accumulates in the kidneys and an increased uptake can also be seen in stomach and gut. The absorbed dose estimation is carried out assuming a multi-exponential elimination from the blood compartment, that is 3 half-times, 6.8 h (40%), 1.28 d (22%) and 19.4d (38%). Any \(^{99m}\text{Tc}\)-HSA outside the blood compartment is assumed to be eliminated quickly through the kidneys. The effective dose has been calculated to 7.8 μSv/MBq. The absorbed dose to the heart wall is estimated to 20 μSv/MBq, spleen 14 μSv/MBq and kidneys 8 μSv/MBq.

**Abbreviation:** HSA = Human serum albumin.

**Related Articles:** Tc-99m-albumin macroaggregates (MAA), Tc-99m-albumin microspheres (HAM)


**Tc-99m-albumin macroaggregates (MAA)**

(Nuclear Medicine) Technetium-99m-macroaggregated albumin, \(^{99m}\text{Tc-MAA}\), is used for lung perfusion scintigraphy for diagnosis of pulmonary emboli. The distribution of MAA is related to regional pulmonary blood flow.

MAA is human serum albumin (the most common protein in the blood) that has been denatured by heat in an agitated aqueous solution to form macroaggregates with particle size of 10–90 μm. A kit contains about 1.5–2 million albumin particles (2 mg) of which 90% have a size distribution of 10–50 μm. Labelling is performed by adding 2–10 mL \(^{99m}\text{Tc}\)-pertechnetate with an activity up to 3.7 GBq. After preparation, the injection solution should be used within 6 h. The radiochemical purity can be tested by paper or thin-layer chromatography (recommended), and the purity should not be less than 90%.

After IV injection 90% of the \(^{99m}\text{Tc}\)-MAA is extracted during the first pass and sequestrated in the lung capillaries (8.2 ± 1.5 μm) and arterioles (25 ± 10 μm). Particles smaller than the capillaries are trapped by the RES in the liver.

The effective dose of IV administered \(^{99m}\text{Tc}\)-MAA is 12 μSv/MBq and the organs that receive highest absorbed dose is lungs (66 μGy/MBq), liver (16 μGy/MBq) and bladder (87 μGy/MBq).

**Abbreviation:** MAA = Macroaggregated albumin.

**Related Articles:** Radiochemical purity, Technetium-99m


**Tc-99m-albumin microcolloid**

Technetium-99m-albumin microcolloids are used for liver and spleen scintigraphy. The time of examination performed is 15–60 min after an IV injection. Common trade names are ALBURES (Solco) and Microlite (DuPont Merck). The kit contains sterile, lyophilised, preformed HSA-microcolloids with a size distribution of 0.2–2.0 μm. Particles smaller than the capillaries are trapped by the RES in the liver.

After IV injection 90% of the \(^{99m}\text{Tc}\)-MAA is extracted during the first pass and sequestrated in the lung capillaries (8.2 ± 1.5 μm) and arterioles (25 ± 10 μm). Particles smaller than the capillaries are trapped by the RES in the liver.

The effective dose of IV administered \(^{99m}\text{Tc}\)-MAA is 12 μSv/MBq and the organs that receive highest absorbed dose is lungs (66 μGy/MBq), liver (16 μGy/MBq) and bladder (87 μGy/MBq).

**Abbreviation:** MAA = Macroaggregated albumin.

**Related Articles:** Radiochemical purity, Technetium-99m


The absorbed dose estimation is carried out assuming a multi-exponential elimination from the blood compartment, that is 3 half-times, 6.8 h (40%), 1.28 d (22%) and 19.4d (38%). Any \(^{99m}\text{Tc}\)-HSA outside the blood compartment is assumed to be eliminated quickly through the kidneys. The effective dose has been calculated to 7.8 μSv/MBq. The absorbed dose to the heart wall is estimated to 20 μSv/MBq, spleen 14 μSv/MBq and kidneys 8 μSv/MBq.

**Abbreviation:** HSA = Human serum albumin.

**Related Articles:** Tc-99m-albumin macroaggregates (MAA), Tc-99m-albumin microspheres (HAM)

most exposed, with an absorbed dose to the liver, spleen and red bone marrow of 71, 75, and 11 μGy/MBq respectively. The effective dose is approximately 10 μSv/MBq. Normally 40–150 MBq is administered IV for planar imaging, whereas 200 MBq is used for SPECT.

**Abbreviations:** IV = Intravenously and RES = Reticuloendotheial system (of the liver).

**Related Articles:** Tc-99m tin colloids, Tc-99m albumin nanocolloid


Tc-99m-albumin millimicrospheres

(Nuclear Medicine) Tc-99m-albumin millimicrospheres (Tc-milli-HAM) is used for liver and spleen scintigraphy, regional liver perfusion and RES function, and bone marrow scintigraphy. In these applications the radiopharmaceutical is administered intravenously. It may also be used for pulmonary ventilation studies when the substance is inhaled as an aerosol (nebulisation). The time of examination is 10–60 min post-injection for liver and spleen imaging and 45–60 min for bone marrow imaging. For pulmonary ventilation scintigraphy is performed immediately after inhalation.

A kit contains sterile, lyophilized, preformed HSA millimicrospheres with a 90% size distribution 0.3–0.8 μm. Tc-pertechnetate is simple added in a volume of 1–5 mL (max 3 GBq) and the reaction is left in room temperature for 15 min. No heating is needed. Commercial kits available are Nanotec (Sorin). For QC of Tc-milli-HAM thin-layer chromatography (TLC) with 85% methanol as solvent is recommended. The radiochemical purity should be at least 95%. The preparation is stable for 6 h.

Small amounts of IV administered Tc-milli-HAM are quickly cleared from the blood in the first minutes. About 85% is phagocytosed by RES with a maximum uptake 5–10 min post-injection, and the retention reflects the function of the Kupffer cells. Plasma clearance rate depends on particle size of colloids.

As for most colloids, the liver, spleen and bone marrow are the most exposed organs. Absorbed doses have been estimated to be 71, 75, and 11 μGy/MBq, respectively. The effective dose is estimated to be 10 μSv/MBq. Normally 40–150 MBq is administered IV for planar imaging, whereas 200 MBq is used for SPECT.

**Abbreviation:** HSA = Human serum albumin.

**Related Articles:** Tc-99m-labelled microcolloids, Tc-99m-albumin microcolloid, Tc-99m-labelled nanocolloids, Tc-99m-albumin nanocolloid


Tc-99m-arcitumomab

(Nuclear Medicine) Tc-99m-arcitumomab is a carcinoembryonic antigen complex, used for scintigraphic studies of malignancies in patients with colorectal carcinoma and with recurrence or metastasis. An example of a commercial product is CEA-Scan (Immunomedics Europe). A Tc-CEA-Scan kit contains lyophilised, sterile 1.25 mg arcitumomab IMMU-4 Fab’ anti-CEA MABf in argon atmosphere, ready for labelling with Tc-pertechnetate.
The radiopharmaceutical is given IV in an activity of 740–1110 MBq (in 2 mL).

The radiochemical purity should be at least 90%, and is tested for free 99mTc-pertechnetate before administration. Thin-layer chromatography (instant (I)TLC-silica gel fibreglass sheets) and ace-tone as solvent is recommended.

99mTc-arcitumomab is excreted by the kidneys; thus, together with the urinary bladder, liver and spleen are the most exposed organs: 89, 10, 8.7 and 14 μGy/MBq respectively. The effective dose is estimated to 11 μSv/MBq.

**Abbreviations:** CEA = Carcinoembryonic antigen and IV = Intravenous.


**Tc-99m; Ceretec**

(Nuclear Medicine) This is 99mTc D.L-HMPAO, hexamethylpropylene amine oxime, also called exametazine. The commercial product is Ceretec (GE Healthcare). For further details see **Tc-99m HMPAO**.

**Tc-99m-diphosphonates (DPD, HDP, MDP, HEDSPA)**

(Nuclear Medicine) 99mTc-diphosphonates are available as different complexes. Phosphonate and phosphate complexes are both taken up in the bone to a high extent. Phosphonates are more stable in vivo than phosphates because the P-O-P bond in the latter is easily broken down by phosphatase enzyme, whereas the P-C-P bond in diphosphonate is not. Thus, 99mTc-diphosphonates can be used for skeletal imaging with great advantages. Examples are diagnosis of primary or metastatic bone tumours, osteomyelitis, localisation of fractures and evaluation of pain in the skeleton in patients with negative x-rays.

99mTc-diphosphonates of frequent occurrence are 1-hydroxyethylidene diphosphonate (HEDP), methylene diphosphonate (MDP), and hydroxymethylene diphosphonate (HDP or HMDP). Commercial kits are available from different manufacturers:

99mTc-DPD (dicarboxypropane diphosphonate) – Teceos (CIS Bio International)

99mTc-HDP (hydroxymethylene diphosphonate) – Technene Scan HDK (Mallinckrodt Medical)

99mTc-MDP (methylene diphosphonate) – MedroGis TCK-14 (CIS Bio International)

99mTc-HEDSPA (hydroxyethylidene diphosphonate) – HEDSPA (Union Carbide)

Independent of the manufacturer, the kit contains lyophilised, sterile, pyrogen-free diphosphonate in nitrogen atmosphere, prepared for easy labelling with 2–10 mL 99mTc – sodium pertechnetate with an activity up to 6.6–18.5 GBq, depending on labelling efficiency, number of patients and scheduled time of administration. After a few minutes at room temperature the preparation is ready for use with an expiry time of 6–8 h. The time of examination for bone imaging is 2 h post-administration.

Historically HEDSPA was the first 99mTc-disphosphonate complex used in early 1970s. MDP was for long the diphosphonate complex of choice. Later both HMDP and HDP were introduced with similar good characteristics, followed by DPD also with excellent merits. The signals for detection of lesions are an increased blood flow to the skeleton and an increased regional uptake in bone metastasis. The detection rate is mainly independent of the diphosphonate complex (DPD, HDP or MDP).

The radiochemical purity can be tested by thin-layer chromatography (recommended), and the purity should not be less than 95%.

After an IV administration 45%–50% of 99mTc-diphosphonates accumulate in the skeleton, while the rest is excreted in the urine. Maximum bone uptake is 1 h post-administration and remains constant for 72 h. The most exposed organs are bone surfaces (63 mGy/MBq), urinary bladder (48 mGy/MBq), red bone marrow (9 mGy/MBq), kidneys and the heart wall (7.3 mGy/MBq). The effective dose is estimated at 6 μSv/MBq.

**Abbreviations:** DPD = Dicarboxypropane diphosphonate, HDP = Hydroxymethylene diphosphonate, HEDSPA = Hydroxyethylidene diphosphonate and MDP = Methylen diphosphonate.

**Related Article:** Tc-99m-pyrophosphate (PYP)


**Tc-99m-DMSA (dimercaptosuccinic acid)**

(Nuclear Medicine) 99mTc-DMSA is used for planar or tomographic imaging of renal cortex function and morphological studies of lesions and areas with reduced function. The radiopharmaceutical can also be used for detection of metastasis of medullary thyroid carcinoma. In the kidney 99mTc-DMSA is retained in the renal parenchyma by tubular fixation and indicates its function. Tumours, cysts and abscesses appear as cold spots. The time of examination for renal imaging is 1–3 h post-IV administration.

Radiochemical purity of 99mTc-DMSA should not be less than 95%, which is tested by thin-layer chromatography (TLC) on silica gel fibre-glass sheets with methyl ethyl ketone (MEK) and saline as solvents.

The uptake of 99mTc-DMSA is related to the renal cortical perfusion. One hour after an IV injection 24% of 99mTc-DMSA has taken up in the renal parenchyma. The complex is excreted with the urine with significant uptake in the bladder as well. Most of the circulating 99mTc-DMSA is loosely bound to plasma proteins, and no other organs show any uptake. Thus, the organs receiving largest absorbed dose are kidneys, bladder wall and adrenals, 180, 18 and 12 μGy/MBq respectively. The effective dose is 8.8 μSv/MBq.

**Abbreviation:** IV = Intravenous.
Related Article: Tc-99m-MAG3

Tc-99m-DTPA
(Nuclear Medicine) Technetium-99m Diethylenetriamine penta-acetic acid, 99mTc-DTPA, together with EDTA is the most common chelating agent used in nuclear medicine pharmacy. DTPA has eight possible complexing sites, three nitrogen and five carboxylic sites. The DTPA-chelate binds to Tc(IV), and other metals very tightly. It can be used as a detoxification agent for lead, for example it chelates the metal and speeds up the excretion through the kidneys.

99mTc-DTPA is efficiently transferred from the blood to the urine and used for renal studies. The complex may also be utilised for regional lung ventilation studies as an aerosol that is inhaled (e.g. TechnetScan DTPA/Aerosol kit). Using an ultrasonic nebulisation barrier leakage.

After IV injection, 99mTc-DTPA passes capillary walls to enter the extravascular space within 4 min. The complex is hydrophilic and negatively charged and thus stays in the extravascular space. The kidneys alone are responsible for the removal of the complex and the renal transit time is 5 min. The plasma clearance is described by a double-exponential function, 99% is excreted with a biological half-life of less than 2 h, the remaining 1% with a half-life of 7 days.

The effective dose of IV administration of 99mTc-DTPA is 5 μSv/MBq and the organ that receive highest absorbed dose is the bladder with 62 μGy/MBq.

Abbreviation: IV = Intravenous injection.
Related Articles: Chelates, Tc-99m

Tc-99m-EC (ethylene dicysteine)
(Nuclear Medicine) 99mTc-EC, ethylene dicysteine, is used for dynamic examination of renal function, determination of the tubular extraction rate, and study of the function of a transplanted kidney. Gamma camera acquisition should start immediately after an IV administration of the 99mTc-EC, and continue for approximately 20 min. Recommended activity administered is 90–120 MBq.

A 99mTc-EC kit consists of 3 vials containing the lyophilised components. The first vial contains ethylene-t, i. dicysteine and ingredients to facilitate labelling with 99mTc-pertechnetate at alkaline pH. Labelling is started by adding Sn2+ as reducing agent from the second vial, and left for 15 min incubation. The third vial contains a buffer as a stabilising agent, which is added to the 99mTc-EC.

The radiochemical control is performed by thin-layer chromatography using a solvent system to separate 99mTc and reduce it from the 99mTc-EC complex. The purity should exceed 95%. The complex is stable for 3–8 h and sufficient for three patients.

The blood clearance of 99mTc-EC complex is rapid, 2–3 min after IV injection no activity is seen in heart and liver. The complex is excreted by tubular secretion in the kidneys. The difference between 99mTc-EC and 99mTc-MAG3 is a higher plasma clearance and lower uptake in liver, bowel and gallbladder for 99mTc-EC.

Kidneys, bladder wall and adrenals are the organs that show highest absorbed doses. For normal kidney function the effective dose is 6.3 μSv/MBq, the absorbed dose to kidney 3.4 μGy/MBq and bladder wall 9.4 μGy/MBq. For reduced kidney function the effective dose is 4.6 μSv/MBq, the absorbed dose to kidney 1.2 μGy/MBq and bladder wall 4.3 μGy/MBq.

Abbreviation: IV = Intravenous.
Related Articles: Tc-99m-MAG3 (mercaptoacetyltriglycine), Tc-99m-DTPA

Tc-99m-ECD
(Nuclear Medicine) 99mTc-ECD, ethyl cysteinate dimer (also called bicisate), is a substance for brain perfusion imaging. Its trade name is Neurolite (Bristol-Myers Squibb). The Neurolite kit consists of 3 vials. The Neurolite kit consists of 3 vials containing the lyophilised components. The first vial contains ethylene-t, i. dicysteine and ingredients to facilitate labelling with 99mTc-pertechnetate at alkaline pH. Labelling is started by adding Sn2+ as reducing agent from the second vial, and left for 15 min incubation. The third vial contains a buffer as a stabilising agent, which is added to the 99mTc-EC.

The radiochemical control is performed by thin-layer chromatography using a solvent system to separate 99mTc and reduce it from the 99mTc-EC complex. The purity should exceed 95%. The complex is stable for 3–8 h and sufficient for three patients.

The blood clearance of 99mTc-EC complex is rapid, 2–3 min after IV injection no activity is seen in heart and liver. The complex is excreted by tubular secretion in the kidneys. The difference between 99mTc-EC and 99mTc-MAG3 is a higher plasma clearance and lower uptake in liver, bowel and gallbladder for 99mTc-EC.

Kidneys, bladder wall and adrenals are the organs that show highest absorbed doses. For normal kidney function the effective dose is 6.3 μSv/MBq, the absorbed dose to kidney 3.4 μGy/MBq and bladder wall 9.4 μGy/MBq. For reduced kidney function the effective dose is 4.6 μSv/MBq, the absorbed dose to kidney 1.2 μGy/MBq and bladder wall 4.3 μGy/MBq.

Abbreviation: IV = Intravenous.
Related Articles: Tc-99m-MAG3 (mercaptoacetyltriglycine), Tc-99m-DTPA


**Tc-99m-HMPAO**

(Nuclear Medicine) 99mTc-D,L-HMPAO, hexamethylpropylene amine oxide, also called exametazime. The commercial product is Ceretec (GE Healthcare). A Ceretec-kit contains lyophilised, sterile, pyrogen-free substance in nitrogen atmosphere, prepared for labelling with 99mTc – sodium pertechnetate. Five microlitres of Na\(^{99m}\)TcO\(_4\) with an activity of 370–1100 MBq is added to the vial, which should be gently inverted for 10s and the reaction allowed to proceed at room temperature for 5 min. A mixture of methylene blue in phosphate buffer may be used for stabilisation of the preparation.

When fresh 99mTc is reduced with Sn\(^{2+}\), the 99mTc D,L-HMPAO complex is formed rapidly. Any oxidant should be avoided. The preparation should normally be used within 1h after labelling. With the stabilizer (methylen blue) the preparation is stable for 4h. 99mTc-HMPAO is used for brain scintigraphy, normally SPECT, for the diagnosis of perfusion defects of regional cerebral blood flow (rCBF); focal perfusion abnormalities, stroke, and multi-infarct and degenerative dementia. The complex can also be used for labelling of leukocytes.

99mTc-exametazime is lipophilic and can cross the blood–brain barrier (BBB) with high efficacy. Due to the in vivo instability of the complex, the secondary 99mTc-HMPAO formed cannot pass the BBB and is trapped inside the brain.

The labelled 99mTc-HMPAO must be tested for radiochemical purity before injected to the patient, and the labelling efficiency should be at least 80%, that is lipophilic 99mTc-HMPAO. It is normally checked using thin-layer chromatography (TLC) on Gelman ITLC silica gel sheets and paper chromatography on Whatman 1 strips, using three solvent systems for the analysis of lipophilic 99mTc-HMPAO, secondary hydrophilic complex, unbound 99mTc and reduced, hydrolyzed 99mTc.

After 5min, the 99mTc-HMPAO has disappeared from the blood and the distribution of activity in the brain represents the true image of the initial blood flow. It remains constant for about 24h, and the elimination rate is very slow (1% per hour).

The effective dose for 99mTc-HMPAO is 9.3 μSv/MBq. The organs most exposed are kidneys (32 μGy/MBq), bladder wall (22 μGy/MBq) and lungs (11 μGy/MBq).

See also Tc-99m ECD, which is a similar radiopharmaceutical used for brain perfusion imaging.

**Abbreviations:** CBF = Cerebral blood flow.

**Related Articles:** Radiopharmaceutical purity, Tc-99m-HMPAO, TLC


**Tc-99m-IDA (iminodiacetic acid)**

(Nuclear Medicine) 99mTc-IDA, iminodiacetic acid, is a hepatobiliary radiopharmaceutical for imaging and evaluation of hepatocyte function, obstruction of the cystic duct, examination of acute cholecystitis, and common bile duct obstruction. For hepatobiliary scintigraphy, 150 MBq of activity is recommended. The examination is started 5min post-IV administration, and the scintigrams are taken at 10min interval for normally 1h.

The 99mTc-IDA kit contains sterile, lyophilised, preformed ingredients. Labelling is simple performed with 1–5 mL 99mTc-pertechnetate (300–1500 MBq) added to the vial. The complex is stable for 6h after preparation. The radiochemical purity is tested with thin-layer chromatography, TLC, with two different solvent systems: (1) using saline for control of unbound 99mTc at the solvent front and the sum of reduced, hydrolysed technetium and the 99mTc-IDA at the start, and (2) using Gelman ITLC-SG and acetonitrile/water as solvent, where reduced, hydrolysed technetium is identified separately at the start. The labelling efficiency should be at least 95%.

Following IV administration 99mTc-IDA is bound to plasma proteins and transported to the liver. In the liver the complex is dissociated to facilitate active transport of the 99mTc-IDA complex into hepatocytes. In patients with normal liver function, the maximum uptake is at 12min, and the gallbladder is clearly visible within 20min post-injection.

The effective dose for 99mTc-IDA is 17 μSv/MBq. The most exposed organs are gallbladder (110 μGy/MBq), the wall of the guts, that is ULI (86 μGy/MBq), LLI (59 μGy/MBq), SI (44 μGy/MBq), and the liver (14 μGy/MBq).

**Abbreviations:** IV = Intravenous, LLI = Lower large intestine, SI = Small intestine and ULI = Upper large intestine.


**Tc-99m-labelled bone imaging agents**

(Nuclear Medicine) See Tc-99m-pyrophosphate (PYP), Tc-99m-diphosphonates

**Tc-99m-labelled colloids**

(Nuclear Medicine) See Tc-99m-labelled microcolloids, Tc-99m-rhenium sulphide colloid, Tc-99m-albumin microcolloid, Tc-99m-albumin millimicrospheres, Tc-99m-labelled nanocolloids, Tc-99m-albumin nanocolloid

**Tc-99m-labelled erythrocytes**

(Nuclear Medicine) Imaging-labelled erythrocytes, or red blood cells (RBC), can be used clinically for radionuclide angiography, for example regional imaging of blood pools, ejection fraction and wall motion, gastrointestinal haemorrhage and blood loss through determination of erythrocyte mass or blood volume. After heat treatment at 45.9°C erythrocytes will get damaged (become denatured) and can be used for spleen scintigraphy.

Erythrocytes can be labelled, using in vitro, in vivo, or modified in vivo methods. Each technique has its own advantage and disadvantage, but since the in vitro technique has best labelling efficiency, 98%, it is often the preferred technique although being more time consuming.

The basic principle of 99mTc-labelling of erythrocytes involves mixing the cells with Sn²⁺ (stannous chloride) ions followed by addition of 99mTc-pertechnetate. The Sn²⁺ enters into the erythrocytes and afterwards 99mTc-pertechnetate diffuses into it. The pertechnetate with oxidation state 7+ is reduced by the Sn²⁺ to a lower oxidation state and nearly 80% binds to the beta-chain of the globulin part of the haemoglobin molecule and the rest to the heme molecule.

For in vivo labelling Sn-PYP kit is diluted with saline and a sufficient volume is injected IV into the patient. After 20–30 min 99mTc-pertechnetate is injected which tags the erythrocytes immediately. Labelling efficiency is 80%–90%. In the modified in vivo method the patient blood is labelled in vivo with Sn-PYP, but blood is redrawn and labelled in a sterile vial with 99mTc-pertechnetate and injected back into the patient following flushing with saline. Labelling efficiency is above 95%.

The most exposed organs are the heart, lungs, kidneys and liver with absorbed doses of 23, 18, 18 and 13 μGy/MBq respectively. The effective dose for 99mTc-labelled erythrocytes is 7 μSv/MBq.

**Abbreivation:** RBC = Red blood cells.

**Related Articles:** Tc-99m-labelled leukocytes, Tc-99m-diphosphonates, Tc-99m-PYP


**Tc-99m-labelled leukocytes**

(Nuclear Medicine) Labelled leukocytes or white blood cells (WBC) can be used for abdominal scintigraphy to locate sites of focal infection, that is abdominal abscess, sepsis, or examination of fever of unknown origin. It can also be used for detection of osteomyelitis in children offering superior information compared with 99mTc-pyrophosphates. Leukocytes can be labelled either with 99mTc or 111In. Imaging is normally performed at 1, 2 and/or 24 h post-administration.

The lipophilic 99mTc-exametazime (HMPAO) can be used to label leukocytes without affecting cell viability. The activity used for abdominal scintigraphy is 185–370 MBq.

For description of the kit and for detailed procedure of the biological labelling of leukocytes, see 99mTc-HMPAO. Published and approved descriptions should carefully be followed. Briefly, labelling of the leukocytes includes withdrawing of the patient’s blood into syringes with ACD-solution, allowing the syringe to stand for 30–40 min in room temperature for erythrocytes to sediment. The leukocyte- and platelet-rich plasma is drawn into a sterile tube for gentle centrifugation. Removal of the supernatant platelet-rich plasma leaves a pellet of mixed leukocytes on the bottom of the tube, in which 1 mL 99mTc-HMPAO is added and the white blood cells are incubated for 10 min. After incubation, cell-free plasma is added and the cell mixture washed by gentle centrifugation. Finally, new cell-free plasma containing ACD is added to the pellet of leukocytes and made ready for re-injection into the patient without delay.

The labelled 99mTc-HMPAO-leukocytes should be tested before reinjection into the patient. The radiochemical purity should be at least 95%, and tested by TLC similar to the control of 99mTc-HMPAO. Free 99mTc-pertechnetate and unbound 99mTc-HMPAO are the impurities to be measured. The viability of the cells is normally not tested.

99mTc-labelled leukocytes show an initial transitory uptake in the lung, followed by an accumulation in liver, spleen and bone marrow. Kidneys and the gall bladder may also be seen. The 99mTc complex is slowly released from the leukocytes, excreted by the urine and faeces.

The absorbed doses for 99mTc-labelled leukocytes are 150 μGy/MBq for the spleen, 20 μGy/MBq for the liver, and 23 μGy/MBq for the red bone marrow. The effective dose is 11 μSv/MBq.

**Abbreviations:** ACD = Acid-citrate-dextrose, IV = Intravenous, RBC = Red blood cells (erythrocytes), TLC = Thin-layer chromatography and WBC = White blood cells (leukocytes).

**Related Article:** Tc-99m-HMPAO


Tc-99m-labelled microcolloids
(Nuclear Medicine) See Tc-99m tin colloids, Tc-99m rhenium sulphide colloid, Tc-99m albumin microcolloid

Tc-99m-labelled monoclonal antibodies
(Nuclear Medicine) See Tc-99m-arcitumomab

Tc-99m-labelled nanocolloids
(Nuclear Medicine) See Tc-99m-rhenium sulphide nanocolloid, Tc-99m-albumin nanocolloid

Tc-99m-labelled red blood cells (RBC)
(Nuclear Medicine) See Tc-99m-labelled erythrocytes

Tc-99m-labelled white blood cells
(Nuclear Medicine) See Tc-99m-labelled leukocytes

Tc-99m-MAG₃ (mercaptoacetyltriglycine)
(Nuclear Medicine) Technetium-99m mercaptoacetyltriglycine, MAG₃, is used for renal imaging, that is renography. Using MAG₃ information may be obtained of renal anatomy and function, for example to demonstrate satisfactory renal perfusion, renal tubular function, and determination of the excretion rate.

The MAG₃ kit contains lyophilised, sterile, pyrogen-free substance in nitrogen atmosphere, prepared for labelling with ⁹⁹mTc – sodium pertechnetate. The commercial product is called Technescan MAG₃, (Mallinckrodt).

Small amounts of ⁹⁹mTc – impurities may be formed during labelling, which is taken up in the liver. The radiochemical purity shall be at least 90% and may be checked using thin-layer chromatography (TLC).

⁹⁹mTc-MAG₃ is quickly distributed in the extracellular fluid and excreted solely by the kidneys after IV injection with a plasma elimination described by two biological half-lives, 3.2 and 16.9 min. The maximal uptake in the renal system is seen after 3–4 min, and 3 h post-injection the blood activity is less than 1%.

Renal clearance is dependant on the function of the kidneys and the urogenital system. The mechanism of excretion is predomi-
nately based on renal tubular secretion, and glomerular filtration rate accounts for less than 2% of the total clearance.

The most exposed organs are the kidneys and bladder wall. Normally the renal transit time is about 4 min. Depending on the function of the kidneys the radiation doses are as follows:

Normal function: Effective dose 7.3 μSv/MBq (kidneys 3.4 μGy/MBq)

Abnormal function: Effective dose 6.3 μSv/MBq (kidneys μGy/MBq)

Acute unilateral blockage: Effective dose 10 μSv/MBq (kidneys 420 μGy/MBq)

Related Articles: Radiochemical purity, Technetium-99m, Tc-99m DTPA

Tc-99m rhenium sulphide colloids

(Nuclear Medicine) Technetium-99m-labelled rhenium sulphide colloids are a sterile, pyrogen-free, light brown solution for IV administration. The radiopharmaceutical is used for liver and spleen imaging, and occasionally for bone marrow imaging. After oral administration it can be used for imaging of digestive transport and transit time, and for gastroduodenal motor activity. The time of examination is 15–20 min post-injection for liver and spleen scintigraphy, 1 h for bone marrow, and for digestive dynamic imaging (transit time) immediately after administration.

A 99mTc-(Re) sulphide colloid kit contains lyophilised, sterile, pyrogen-free, substance in nitrogen atmosphere, prepared for labelling with 99mTc – sodium pertechnetate. The basic principle for preparation of 99mTc –(Re) sulphide colloid is in principle the same as for 99mTc- sulphur colloid, that is colloidal sulphur and technetium heptasulphide are formed at acidic pH in a water bath at 100°C. Commercial kits are HepatoCis (TCK-1) and Sulfotec Sorin.

The particle size distribution is 0.3 and 0.8 μm. Paper chromatography using saline as solvent is recommended for the radioclinical control (yield) of the preparation. Labelling efficiency in a good preparation should not be less than 92%. Factors affecting the yield are mostly related to pH, wrong mixing order, low heating temperature, too large volume at heating, incorrect boiling time. The 99mTc-(Re) sulphide colloid kit is normally stable for 6 h.

With a particle size of 0.3–0.8 μm, 80%–90% are rapidly removed from the blood and accumulated in the liver by phagocytosis (RES), 4%–8% in the spleen, and 3%–5% in the bone marrow. Larger colloids show increased uptake in the spleen, while smaller colloidal particles localize in the bone marrow.

The most exposed organs are the spleen, liver and red bone marrow, which result in absorbed doses of 75, 71 and 11 μGy/MBq respectively. The effective dose is approximately 10 μSv/MBq. Normally 75–150 MBq administered IV for planar imaging, whereas 200 MBq is used for SPECT.

Related Articles: Tc-99m sulphur colloids, Tc-99m tin colloids, Tc-99m albumin microlloids, Stannous chloride


Related Articles: Tc-99m nanocolloid, Tc-99m sulphur colloid

Tc-99m-rhenium sulphide nanocolloid

(Nuclear Medicine) Technetium-99m-labelled rhenium sulphide nanocolloid is used for imaging of lymphoscintigraphy, imaging of lymphatic flow and regional lymph nodes in torso and extremities after subcutaneous or interstitial administration. It is also used for sentinel lymph node (SLN) scintigraphy in which the radiopharmaceutical is administered subdermally or peritumourally. Imaging of the SLN can detect the first lymph node of a primary tumour before surgery. After oral administration, 99mTc-rhenium sulphide nanocolloid is used for imaging of motility disorder or the esophagus or gastroduodenal motor function.

99mTc-rhenium sulphide nanocolloid is commercially available as NanoCis TCK-17 (CIS Bio) or Lymphoscint Solco (GE Healthcare). A preformed 99mTc-kit consists of two vials. One contains sterile, pyrogen-free solution of rhenium sulphide and other ingredients. The second vial contains sodium pyrophosphate and stannous chloride. The two constituents are mixed and 99mTc-pertechnetate is added. The mixture is placed in a boiling water bath for 15–30 min. After cooling, the radiopharmaceutical is ready for use.

Several factors affect the labelling efficiency, that is low colloid formation and thus low specific activity are principally related to pH, wrong mixing order, low heating temperature, too large volume at heating, incorrect boiling time, or a failing kit. The 99mTc-(Re) nanocolloid kit is normally stable for 4 h after preparation. 99mTc-(Re)-sulphide nanocolloid labelling efficiency can be tested using paper chromatography, for example. Whatman 31 paper using methyl ethyl ketone (MEK) as solvent. TLC may also be used for quality control in which acetone is used as a solvent. The radiochemical purity should be at least 95%.

After subcutaneous administration 99mTc-(Re)-sulphide nanocolloid is transported with the interstitial fluid through the lymphatic capillaries into the lymph ducts, and retained by the regional lymph nodes. Clearance is slow and depends on the motion of the patients’ extremities.

Dosimetric evaluation after subcutaneous administration is based on the assumption that about 5%–15% of the activity is distributed among 10–20 lymph nodes. The highest absorbed dose may be received at the site of injection and could be 10–20 μGy/MBq and to a 5 g lymph node 0.5–0.7 mGy/MBq. The effective dose has been estimated to be approximately 5 μSv/MBq.

Abbreviation: SC = Subcutaneous.

Related Articles: Tc-99m nanocolloid, Tc-99m sulphur colloid


Tc-99m-sodium pertechnetate

(Nuclear Medicine) Chemical name: Sodium pertechnetate 99mTc

Abbreviated name: Na99mTcO4

Examples of commercial products: 99Mo/99mTc-generators; Ultra-Teknekov® DTE (COVIDEN – Mallinckrodt); DRYTEC (GE Healthcare); TechneLite® (Bristol-Myers Squibb).

Sodium pertechnetate, Na99mTcO4 is eluted carrier-free from a 99Mo/99mTc -generator with sterile 0.9% saline solution in a volume of about 10–20 mL. The activity of the elute depends on the size of the generator, that is the 99Mo column activity, which varies...
normally between 37 GBq (1 Ci) and 370 GBq (10 Ci) at the day of reference. Since the activity of the 99mTc elute relates to the time between elutions, daily elution of the generator at an interval of 24 h will give the best quality of the elute.

Sodium pertechnetate is a highly water soluble product but when reduced it loses its water solubility and forms relatively stable bonds to chelates, proteins and cells. When it is reduced several redox states are possible, making the technetium chemistry quite complex, but at the same time many possibilities to label to different substances.

The oxidation state of TcO₄⁻ is +VII and it is the most stable form of Tc in water and air. The radiopharmaceutical chemistry involves the reduction of pertechnetate to an oxidation state between +VI and +III for labelling of different compounds. The reducing agent of choice is Sn²⁺ (see Stannous chloride).

Sodium pertechnetate, ⁹⁹mTc(+VII) may be administered intravenously as it is for scintigraphy of the thyroid, salivary glands, gastrointestinal tract, that is the walls, (0.16–0.42 mGy/MBq) and the bladder (0.018 mGy/MBq). The effective dose is 0.013 mSv/MBq. If a blocking agent (stable iodine) has been given the absorbed doses will reduce a factor of 10, and the effective dose to 0.0042 mSv/MBq.

Abbreviation: IV = Intravenously.

Related Articles: Radionuclide generator, Stannous chloride, Kits, Blocking agent


**Tc-99m-Sulesomab**

(Nuclear Medicine) ⁹⁹mTc-Sulesomab is antigranulocyte monoclonal antibody fragment complex, used for scintigraphic studies of infection and inflammation in patients with suspected osteomyelitis, joint infection involving implants, inflammatory bowel disease, and foot ulcers in diabetic patients. An available commercial kit is LeukoScan from Immunomedics Europe.

A LeukoScan kit contains lyophilised, sterile sulesomab (IMMU-MN3 Fab²-SH antigranulocyte antibody fragments) in a nitrogen or argon atmosphere, ready for labelling with 1100 MBq ⁹⁹mTc-pertechnetate. The radiopharmaceutical is given IV for planar or SPECT imaging.

The radiochemical purity should be at least 90%, and is tested for free ⁹⁹mTc-pertechnetate before administration. Thin-layer chromatography (instant (I) TLC-silica gel fibreglass sheets) and acetone as solvent is recommended.

⁹⁹mTc-Sulesomab is excreted by the kidneys; thus, they are the most exposed organs together with the urinary bladder, liver and spleen: 89, 10, 8.7 and 14 μSv/MBq respectively. The effective dose is estimated to be 11 μSv/MBq.

Abbreviation: IV = Intravenously.


**Tc-99m-sulphur colloid**

(Nuclear Medicine) Technetium-99m-labelled sulphur colloid is a classical radiopharmaceutical for liver and spleen imaging, and occasionally for bone marrow imaging. It can also be used for scintigraphy of GI blood loss and for gastric emptying using ⁹⁹mTc-labelled scrambled eggs. Today, this radiopharmaceutical is rarely used.

The basic principle of ⁹⁹mTc sulphur colloid labelling is to add an acid to a mixture of ⁹⁹mTc-pertechnetate and sodium thiosulphate and then heat it at 95°C–100°C in a water bath for 5–10 min. The pH (6–7) is adjusted with a suitable buffer. Labelling efficiency is normally close to 0%. Commercial kits are available.

The particle size is 0.1–1 μm with a mean of 0.3 μm. Using a membrane filter (0.1 or 0.2 μm), small particle size sulphur colloid may be used for lymphoscintigraphy as well.

The most exposed organs are the liver, spleen and red bone marrow, which result in absorbed doses of 71, 75 and 11 μGy/MBq respectively. The effective dose is approximately 10 μSv/MBq. Normally 40–150 MBq is administered IV for planar imaging, whereas 200 MBq is used for SPECT.

Related Articles: Tc-99m tin colloids, Tc-99m albumin micro-colloids, Stannous chloride


**Tc-99m-tin colloids**

(Nuclear Medicine) Technetium-99m-labelled tin colloids are in the range of 0.2–0.8 μm particle size. The radiopharmaceutical is used for liver and spleen scintigraphy. IV injected ⁹⁹mTc tin colloids
localize within the liver due to phagocytosis of the reticuloendothelial system (RES). The time of examination is performed 10–15 min after an IV injection. Common trade names are Amerscan Heptate II (GE Healthcare) and Livoscint (Bristol-Myers Squibb).

A kit contains sterile, lyophilised, preformed tin-II-fluoride with a size distribution of 0.2–0.8 μm, sodium fluoride and a stabilising agent (Poloxamer 188). Preformed tin colloid is easily labelled with reduced 99mTc. Maximum activity is 3.7 GBq per vial. The radiochemical purity should at least be 95%, for example tested with TLC.

With a particle size of 0.2–0.8 μm, 80%–90% are rapidly removed from the blood and accumulated in the liver during phagocytosis, 5%–10% in the spleen, and 5%–9% in the bone marrow. These organs are the most exposed, with an approximate absorbed dose of 71, 75 and 11 μGy/MBq respectively. The effective dose is approximately 10 μSv/MBq. Normally 75–185 μBq is administered IV for planar imaging, and 200 MBq for SPECT imaging.

**Abbreviation:** IV = Intravenously.

**Related Articles:** Tc-99m albumin microcolloids, Tc-99m albumin nanocolloid, Stannous chloride


**TD (time delay)**

(Magnetic Resonance) Time delay (TD, also called ‘intersegment delay’) is a delay introduced between successive slice acquisitions in multislice scanning (or between successive partitions in 3D scanning) in some fast imaging techniques.

For example, in turboFLASH imaging the sequence begins with a preparatory, non-selective 180° inversion pulse. Slice acquisition then follows, either in a single shot, filling all of k-space, or in segments, filling many lines of k-space. The process then repeats, with a preparatory pulse and filling of the next slice (or segment). To allow adequate longitudinal recovery after each inverting pulse, a delay TD is introduced from the end of each slice or segment acquisition to the acquisition of the next. Similarly TD is implemented in MP-RAGE, the 3D variant of turboFLASH (Figure T.7).

TD contributes to overall scan time. In turboFLASH the scan time is as follows:

\[
\text{Scan time} = \text{No. of slice} \times (\text{TI No. of lines in phase encode direction} \times \text{TR} + \text{TD})
\]

Note that in MR literature the term time delay (TD) is used quite arbitrarily to denote a delay between events in an acquisition or sequence and its meaning is not limited to the definition given here.


**TE (echo time)**

(Magnetic Resonance) See Echo time (TE)

**Technetium**

(Nuclear Medicine) Technetium is a chemical element with atomic number 43 and is commonly used in medicine (in particular 99mTc) because of its favourable physical characteristics. Technetium does not occur naturally in nature since none of the isotopes are stable. The presence of technetium was predicted by Dmitri Mendeleev because of the gap in the periodic system. The most commonly used technetium isotope in medical imaging is 99mTc because of its favourable physical decay properties:

- Low radiation dose per decay
- Convenient half-life (6.01 h)
- γ-rays are suitable for imaging (140 keV)

99mTc can be labelled to different compounds that target specific physiological bio-processes and it is used to study the functionality of a number of organs and bio-systems, for example brain, myocardium, thyroid, lungs, liver, gallbladder, kidneys and skeleton. 99mTc can also be used to locate tumours.

**Technetium-99m [99Tcm]**

(Nuclear Medicine) Element: technetium

Isotopes: 45 < N < 69

Atomic number (Z): 43

Neutron number (N): 56

Symbol: 99mTc

Production: Generator: 98Mo(n,f) → 99Mo → 99mTc

Mother: 99Mo (reactor/fission produced)

Daughter: 99Tc

Half-life: 6.01 h

![FIGURE T.7](TD in a multislice, single shot turboFLASH sequence.)
Technetium generator

Decay mode: IT
Radiation: gamma, internal conversion electrons, Auger electrons, characteristic x-ray photons
Gamma energy: 140.5 keV (89.1%)
Dose rate from 1 MBq: 0.26 μSv/h at 1 cm (point source); 0.022 μSv/h at 30 cm (10 mL vial)
Absorption (HVL): 0.27 mm lead
Absorption (range of electrons): 0.2 mm glass
Biological half-life: 3–5 h
Critical organ: thyroid gland, gastrointestinal walls
ALL\textsubscript{0\%} (50 mSv): 3000 MBq
Effective dose: 0.020 mSv/MBq (oral); 0.016 mSv/MBq (inhalation)

Clinical Applications: Technetium-99m was introduced in the early 1960s and is nowadays used in over 90% of all nuclear medicine examinations. It is very simply labelled to a number of pre-fabricated kits. 99m\textsubscript{Tc} has advantageous decay characteristics for imaging with scintillation cameras, as well as for a radiation dosimetry point of view.

Related Articles: Technetium-99m-pertechnetate, Tc-99m-generator, Kits

Technetium generator

(Nuclear Medicine) A radionuclide generator with the 99Mo -- 99m\textsubscript{Tc} parent daughter couple. 99m\textsubscript{Tc} is used worldwide in many clinical radionuclide imaging applications. The parent 99Mo is in the form of the molybdate ion, MoO\textsubscript{4}\textsuperscript{2--} which is bound to an alumina column. 99m\textsubscript{Tc} is not as strongly bound to the column and it is therefore possible to elute 99m\textsubscript{Tc} from the column. The elute contains 99m\textsubscript{TcO}\textsubscript{4} (pertechnetate) and normal saline. The saline is used to filter the column. As much as 75%–85% of the available 99m\textsubscript{Tc} is extracted in a single elution. Maximum activity is available again some 24 h after the previous elution when a new 99Mo -- 99m\textsubscript{Tc} equilibrium is reached; but it is also possible to elute usable quantities after 3–6 h. The generator must be recharged on a weekly basis because of the parent decay.

A small fraction of 99Mo will always be eluted with the 99m\textsubscript{Tc}. This is called 99Mo breakthrough. The radiation dose per disintegration for 99Mo is higher than for 99m\textsubscript{Tc} and there is no γ-radiation suitable for imaging from the 99Mo decay so in regard to patient safety the fraction of 99Mo should be kept to a minimum.

Related Article: Molybdenum breakthrough


Technique projections

(General) There is a convention where the radiographic technique projection is identified by the direction of the x-ray beam. For example, for the chest x-ray the projection can either be posterior-anterior or antero-posterior.

Antero-Posterior Projection (AP): The x-ray tube produces an x-ray beam which passes through the back to the front of the patient to produce an image.

Posterior-Anterior Projection (PA): The x-ray tube produces an x-ray beam which passes through the front to the back of the patient to produce an image.

Techniques to improve radionuclide uptake in tumour cells

(Nuclear Medicine) These are techniques that intend to maximise the accumulation of the targeting agent in tumour cells.

There are a number of approaches suggested on how to increase the radionuclide uptake in tumour cells, four of them being

1. Changes of receptor and antigen expression
2. Increased binding affinity
3. Increased cellular retention and internalisation
4. Nuclear localisation

The first approach aims to up- or down-regulate receptors and antigens on the cell surface by using substances like cytokines, hormones or other biological response modifiers. By regulating the receptors and antigens it is possible to improve the target agent uptake.

The second approach aims to increase the binding affinity of the target agent to disseminated tumour cells. If one increases the binding affinity in bulky tumours this might lead to an uneven distribution in the tumour and the method is therefore mainly considered for locating and eliminating disseminated cells.

The third approach aims to increase the retention and internalisation of the target agent. The longer the radion-labelled tracer stays in or near the targeted cell the greater is the radiation dose given to the tumour cell and subsequently greater chance of successful therapy results. In the case of internalisation, the targeting agent is situated closer to the most radiosensitive part of the cell, namely the nuclear DNA. However there may be other problems arising with
internalisation, for example a quicker degeneration of the targeting agent followed by a fast diffusion and clearance of the radionuclide. One alternative to prevent intracellular degradation of the targeting agent is dextranisation.

Nuclear localisation aims to trap the targeting agent inside the cell nucleus, close to the radiosensitive DNA. If possible nuclear localisation therapy would allow the use of much lower activities of β- and α-emitters since the therapeutic effect per decay increases. One principle of nuclear localisation is the use of radio-labelled steroids which binds to steroid-receptor-rich tumour cells. The downside is that conventional steroids have a low residence time inside the cell nucleus, hence not producing the high therapeutic effect desired.

Related Articles: Radionuclide uptake in tumor cells, Extracorporeal elimination


Technologist
(Nuclear Medicine) Technologist is a profession for people educated in the field of technology. In some countries technologist is a certified profession and only students who have graduated from an accredited institution/university are certified.

Teleradiology
(Diagnostic Radiology) Teleradiology is the practice of transmission, usually via the internet, of radiological images for remote reporting or as part of a consultation between radiologists for the purposes of making a diagnosis and producing a report. Images stored electronically, usually within a PACS system, can be transmitted almost instantaneously to another remote image handling system or software package from where they can be reported. This allows for radiological imaging to operate in a small or remote location, even without the presence of a radiologist, whilst still having the capability of providing a relatively prompt report turn around with access to a wider pool of trained and often specialist radiologists only available in larger populated areas. International arrangements could be made to lower the costs of radiological reporting out of hours by sending images to radiologists in different time zones rather than employing radiologists to work outside of normal hours.

The specialist software packages allow for image handling and manipulation and visualisation of images from a number of different specialities and to produce verbally recorded or written reports, possible using voice recognition tools, which can be relayed back to the referring centre.

As teleradiology refers to electronic transfer of radiographs, it can be confused with a specific method of x-ray radiography – Teleradiography.

Teleradiography is a radiographic technique where the distance from the x-ray tube to the detector is about 2 m. The advantages of this geometrical solution are

- Reduction of the penumbra due to the focal spot dimension
- Minimisation of the geometric distortion
- Almost total elimination of the ‘Heel effect’

The overall effect on image quality is improved radiographic sharpness. Teleradiography is often used for chest radiography and orthodontics.

Teletherapy
(Radiotherapy) Radiotherapy, radiation therapy, is the treatment of a disease with ionising radiation.

Depending on the distance between the radiation source and the target volume, that is the tissues to be treated, radiotherapy is divided into two categories: teletherapy and brachytherapy.

- In teletherapy the source is far from the target (Greek word tele, distant, far away).
- In brachytherapy the source is placed close to or inside the target (Greek word brachys, short).

External beam therapy is ordinarily regarded to be identical with teletherapy.

It is to be noted though, that there are external beam techniques with very short source-skin distances, such as contact therapy using kilovoltage x-ray equipment.

Related Articles: External beam therapy, Radiotherapy equipment, Dosimetry, Treatment planning

Temperature coefficient
(Radiation Protection) See Pressure and temperature correction factor

Temperature control
(General) Temperature control is common both within equipment and in the local environment. Temperature control requires some form of sensor – either a simple on–off device, a thermostat, or more complex sensors or probes which have an output proportional to temperature.

On–Off Control: The simpler devices use a heater or cooler wired via a thermostat which turns them on/off when a preset temperature is reached. Thermostats have inbuilt hysteresis or a ‘dead band’ so that, once switched, they require a significant difference in temperature before they will switch back. This prevents any race condition where the thermostat and cooler/heater can reach an equilibrium temperature where the thermostat would fluctuate on and off, and could result in damage to the heater/cooler.

Proportional Control: More complex temperature control systems use sensors which can measure the ambient temperature and coolers/heaters which can either provide varying thermal output or be switched on/off frequently. A microprocessor or analogue filter circuit is used to amplify the sensor signal, compare it with the required temperature value and drive the thermal output in the most efficient manner to minimise the difference. Proportional controls need to be used where thermal tolerances are low or large variations in temperature must be corrected in the shortest time.

Control of Temperature in X-Ray Generators: Temperature control of the anode of an x-ray tube is very important for the correct and safe use of an x-ray generator. The whole tube is immersed in oil (in the tube housing). The simplest temperature control uses a rubber membrane which opens a switch (i.e. interrupts the power supply) when the oil expands due to high temperature of the tube. The most sophisticated one is a microprocessor system which models the tube temperature and controls the exposure parameters (kV, mA, length of exposure) according to the tube thermal load characteristic.

Related Articles: Thermostat, Thermal probe, x-ray generator, Temperature probe

Temperature probe
(General) A general term for any temperature sensor mounted in a long thin housing.

Depending on the temperature range to be monitored, the environment in which it will be used, and the accuracy required, a large number of options are available, among which are as follows:
• Thermocouple – It uses wires of dissimilar metals which, if connected to form a pair of junctions in series, will generate an electro chemical potential across the pair related to the temperature difference (small signal, wide operating range).
• Thermostat – A mechanical switching device which uses the thermal expansion of some material to toggle an electric switch when a preset temperature is reached (usually range −40°C to +150°C).
• Thermistor – A temperature dependent resistor. Resistance usually drops as temperature increases. Must be calibrated and only linear over a short temperature range. Usually range within band and −20°C to +100°C.
• Diode – All semiconductor diodes have a precise negative temperature coefficient of current when reverse biased. (~−20°C to +140°C).
• Optical fibre – Various designs use the reflection of light from the tip of a fibre optic cable as a method of safe monitoring temperature in high magnetic, radiation or electric fields. Among temperature sensitive properties that tips may have are thermo-optical, and thermo-mechanical.
• IR Optical sensor – Non-contacting sensor relying on infra-red spectrum emitted from object being measured. Often requires input of some emissive property of object being measured.

Related Articles: Thermostat, Fluoroptic probe

Temperature sensor (General) See Temperature probe

Temporal filtering (Ultrasound) Temporal filtering is a general term in imaging whereby time-changing features in a signal or image are used to improve image resolution and feature detection. An example of this is recursive filtering in ultrasound images, where frame averaging is used to reduce the effect of speckle. Another implementation is the subtraction of successive images such that images of stationary images are removed and images from moving targets, most commonly blood, are emphasized.

Related Articles: Persistence, Colour flow imaging, Doppler ultrasound

Temporal (instantaneous) peak intensity $I_{TP}$ or $I_{P}$ (Ultrasound) This is the maximum instantaneous intensity. It is the highest intensity to be found anywhere within an ultrasound field. It is determined directly from $p_i$ or $p_s$, whichever has the largest value. It is usually found within a pulse at a transmit focal point.

Related Articles: Intensity, Focal point, Spatial peak intensity, Pressure parameters

Temporal resolution (Ultrasound) The term temporal resolution refers to the frequency with which the image is updated. It applies to B-mode, colour flow and spectral Doppler imaging.

For B-mode and colour flow imaging, the image is updated at a frame rate indicated on the screen. Each individual frame takes a finite time so that an individual image is not an instantaneous snapshot of the area under investigation. This may be important for fast moving structures such as the foetal heart and rapidly changing flow where different parts of the image show different parts of the cardiac cycle. Temporal resolution is dependent on the size of the image and the line density; there is often a compromise to be made between temporal and spatial resolution, especially in colour flow imaging.

For spectral Doppler, the temporal resolution in the sonogram is several hundred Hz, dependent on the processing used, the PRF used and whether B-mode and colour flow is deployed concurrently.

Related Articles: Frame rate, Spatial resolution, Resolution

Temporary implant (Radiotherapy, Brachytherapy) There are two different types of brachytherapy, considering the duration of the treatment:

1. Permanent implants
   a. The sources are placed in the target volume by an interstitial technique.
   b. The sources are then left permanently in the target to decay.
   c. The total dose is delivered over a long time with decreasing dose rate.
   d. There is just ONE implant procedure for the whole treatment.
   e. This is a low dose rate technique.

2. Temporary implants
   a. The sources are placed in the target volume by either interstitial or intracavitary techniques.
   b. The sources stay in the target volume until the correct dose is delivered.
   c. Fixation of applicators can be used for the short HDR treatments, with good control of both total dose and dose distribution.
   d. The treatment is often delivered in several fractions, that is several implant procedures are required.
   e. Temporary implants are suitable for all dose rates, high dose rate, low dose rate and pulsed dose rate.

Temporary implants are often used as a boost treatment together with external beam radiotherapy. A combination of brachytherapy and external beam radiotherapy is for instance the ‘standard radical radiotherapy’ for cervical cancer (see Intracavitary brachytherapy).

Temporary high dose rate interstitial implants can be used also for the treatment of prostate tumours, either as brachytherapy alone or as a boost together with external beam radiotherapy (mostly for the ‘higher risk’ tumours). The basic procedure is similar to the permanent prostate implant; needles are placed in the prostate according to a planned pattern under transrectal ultrasound guidance, using a template to position the interstitial needles (Figures T.8 and T.9).

A remotely controlled afterloading unit with a high intensity $^{192}$Ir-source is used for the treatment. Fluoroscopy is used to position the needles to the correct depth, see Figure T.8. Marker wires are inserted into three of the needles, indicating stop positions for the source in the applicators.

Related Articles: Brachytherapy, Intracavitary brachytherapy, Interstitial brachytherapy, Remote afterloading, Permanent implant, Iridium-192

Tensor (Magnetic Resonance) A tensor is a mathematical idealisation of a geometric or physical quantity that is invariant to transformations, such as rotations. This allows the writing of equations
Tenth value layer (TVL)  
(General) The tenth value layer (TVL) is the thickness of a material required to reduce the intensity of a particular type and quality of radiation to one-tenth. It is expressed in units of distance referred to the specific material (e.g. mm Al). It is used as an alternative to the half value layer (HVL) to indicate the quality or penetrating ability of the radiation. The HVL and TVL are proportional to both the penetrating ability of the radiation and the attenuation coefficient of the material.

Related Articles:  Half value layer (HVL), Beam quality, Beam energy

Terbium  
(General)  
Symbol Tb  
Element category Transition metal  
Mass number A 159  
Atomic number Z 65  
Atomic weight 158.92 g/mol  
Electronic configuration 1s² 2s² 2p⁶ 3s² 3p⁶ 3d¹⁰ 4s² 4p⁶ 4d¹⁰ 4f⁹ 5s² 5p⁶ 6s²  
Melting point 1629 K  
Boiling point 3503 K  
Density near room temperature 8.2 g/cm³  

History: Terbium was discovered in 1843 by Carl Gustaf Mosander as one component of the mineral gadolinite. It is currently obtained mainly from monazite sand using an ion exchange process which yields a mixture rich in lanthanide elements. It is used as a doping agent for certain types of solid-state devices and as a stabiliser in high temperature fuel cells.

Medical Applications: X-ray phosphor – rare-earth phosphors such as terbium-activated gadolinium oxysulphide are sometimes used as image intensifying screens in x-ray imaging. Such phosphors luminesce in green and have a relatively high rate of conversion between absorbed x-rays and emitted visible photons.

Related Articles: Phosphor, Phosphorescence, X-ray image intensifier

TERMA  
(Radiotherapy) TERMA is an acronym for total energy released per unit of mass and represents the energy that is imparted to the secondary charged particles and retained by the scatter photon when the primary photons interact in a unit mass. The TERMA distribution differential in energy $T_E(r)$, that is the total radiant energy released per unit mass by primary photons of energy $E$ with energy fluence $\Psi_E(r)$ in a medium of density $\rho(r)$ is given by

$$T_E(r) = (r/r_0)^2 \frac{\mu(E,r)}{\rho(r)} \Psi_E(r_0) \exp\left(-\int_{s_0}^r \mu(E,l)dl\right)$$

where

$\mu(E,r)$ is the linear attenuation coefficient of the medium at $r$

$\Psi_E(r_0)$ is the energy fluence differential in energy on a reference plane on which the ray from the source to $r$ intersects at $r_0$

The TERMA, which can be thought of as the energy lost out by the primary photon beam in a unit mass, is always larger than the Kerma by a factor $\mu/\mu_\lambda$.  

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**FIGURE T.8** Temporary interactive prostate implant showing all needles placed in the prostate, the ultrasound transducer and the Foley catheter in the bladder with contrast in the balloon, marker wires in three needles.

**FIGURE T.9** Template for HDR needles with locking mechanism (double template system; screw on top displacing one template) to hold all the needles stably in place during the treatment.
The TERMA is generally used as a factor that modulates the energy deposition kernel in the convolution method to calculate the photon beam dose distribution.

The dose in a homogenous phantom is obtained by convolving the TERMA distribution with the kernels. The convolution integral assumes that the kernel is spatially invariant and the integration is carried out over all space, although in practice the integration need only be done in regions where the product of TERMA and photon density is reasonable large.

If the total energy imparted to a unit of mass at \( r \) is the TERMA \( T(r') \), the energy deposited in a unit volume at another point \( r \) due to this energy imparted is given by \( T(r')H(r-r') \) where \( H(r-r') \) is the kernel function for a displacement \( r-r' \) from the kernel origin. The kernel value can be separated into its primary and scattered components, respectively \( H_p \) and \( H_s \). The total dose at \( r \) is given by integrating over all unit masses in the irradiated volume. The dose at \( r \) is therefore given by

\[
D(r) = \int T(r')\left[H_p(r-r') + H_s(r-r')\right] dr^3
\]

**Terminal voltage**

*(General)* A term describing the potential difference measured between two specified ‘terminals’ or connections to the output of a power supply or electrical circuit. Where only one terminal is specified, the second is assumed to be the reference zero voltage point, commonly the chassis or ‘ground’ potential.

Usually the terminal voltage is specified for a particular situation such as ‘no load’, ‘open circuit’ or ‘full load’ state, so that a check can be made against some operational specification.

**Tertiary collimator**

*(Radiotherapy)* This is an additional collimator beyond the main jaws (secondary collimators) on the beam path towards the patient. The tertiary collimator can be permanently mounted within the head of the linac (e.g. MLCs) or externally mounted as and when required (e.g. micro-MLCs).

**Related Articles:** Collimation, Collimator, Multileaf collimator, Secondary collimator, Treatment head, Conformal radiotherapy

**Tesla**

*(Magnetic Resonance)* Tesla is the SI (Système International) unit used to express the strength of a magnetic field. In everyday life, 1 T is a very large unit. The typical strength of the earth’s magnetic field is 50μT, and fields in the mT range are relatively rare. However, the magnets used in MRI have static fields with flux densities of the order of tesla. The switched gradient field used for imaging have magnets used in MRI have static fields with flux densities of the order of tesla. The switched gradient field used for imaging have transport magnets of the order of tesla. The switched gradient field used for imaging have magnets used in MRI have static fields with flux densities of the order of tesla.

The tesla is named in honour of Nikola Tesla (1856–1943), a Serbian-American electrical engineer who invented numerous devices and laid the foundations for the electrical power industry.

**Test voltage**

*(General)* A term describing the potential difference to be provided to the input of a device or circuit section in order that further measurements may be made on the device with a known input state. Such a test voltage must usually be provided by an external voltage source such as a DC power supply.

In some situations the test voltage will represent a specific voltage to be selected by the operator from a range of voltages already available within a device under test (e.g. the kV setting for an x-ray exposure).

**Texture**

*(General)* This is a term used in image processing. An object can have a smooth or coarse texture depending on its spatial frequency content. A smooth object has only low frequency components whereas a coarse object has high frequency components. A high-pass filter or a Fourier transform operation will identify the spatial frequencies of the objects within an image.

**TFD (target film distance)**

*(Diagnostic Radiology)* See Target film distance (TFD)

**Thallium bromide**

*(General)*

<table>
<thead>
<tr>
<th>Symbol</th>
<th>TIBr</th>
</tr>
</thead>
<tbody>
<tr>
<td>Family</td>
<td>Metal halide</td>
</tr>
<tr>
<td>Molecular weight</td>
<td>284.31</td>
</tr>
<tr>
<td>Boiling point</td>
<td>Around 1088 K</td>
</tr>
<tr>
<td>Melting point</td>
<td>Around 753 K</td>
</tr>
<tr>
<td>Density</td>
<td>7.5 × 10^3 g/m^3</td>
</tr>
</tbody>
</table>

**Background:** Thallium bromide is a crystalline semiconductor made of thallium and bromine atoms. Although its use has previously been limited due to relatively poor chemical purity, new purification methods have improved gamma and x-ray detection performance.

**Manufacture:** Crystals of thallium bromide can be grown using the travelling solvent floating zone technique, whereby sections of a mixed thallium and bromine rod are melted using focused infrared heat as the rod travels through the focus. It can also be formed by the cooling of molten thallium bromide from a region containing a seed crystal, a method known as the Bridgman–Stockbarger technique.

**Use in Medicine:** Ionising radiation detectors – Thallium bromide has a high atomic number (Z = 81 and Z = 35 for thallium and bromine, respectively) and a high radiation stopping power. This property, combined with a wide band gap of 2.68 eV, makes it useful as an x- and gamma-ray photon detector, which has low noise at room temperature.

Thallium bromide semiconductor crystals can be used as a detection material in digital radiography applications, and thin films of the material may also have a role in xeroradiography, whereby an x-ray image is printed onto paper with toner which has been attracted to the charges created by excitation of semiconductor electrons.

Infrared detectors – Thallium bromide and thallium iodide can be combined to form a crystal (thallium bromide-iodide) that performs as an infrared radiation detector. Due to its transmission of infrared light up to wavelengths of around 50 μm, it is also commonly found in optical lenses and other optical instruments.

**The European pharmacopeia**

*(Nuclear Medicine)* The European pharmacopeia (Greek: pharmakon = drug and poieo = to work or manufacture) is an official
The 5 Rs of radiobiology

(Radiotherapy) The 5 Rs of radiobiology are the biological factors that influence the response of tumours and normal tissues to radiation and account for the efficacy of fractionated treatment. They are radiosensitivity, repair, repopulation, redistribution and reoxygenation.

The basis of fractionation in radiotherapy can be understood by consideration of these factors. Dividing the radiotherapy dose into fractions spares normal tissues since the cells can repair some of the radiation damage in the time between fractions and cell repopulation will occur provided the overall time is sufficiently long. Conversely, fractionating the dose increases the damage to the tumour due to reoxygenation and the redistribution of cells into radiosensitive phases of the cell cycle between fractions.

Generally, repair and repopulation will tend to make the tissue more resistant to a second dose of radiation; redistribution and reoxygenation tend to make it more sensitive. Although the repopulation of normal tissues is a benefit of fractionation, it should be noted that excessive prolongation of the overall treatment time can adversely affect local tumour control due to tumour cell repopulation. This has been observed in a number of tumour sites including squamous cell cancers of the head and neck region, non-small cell lung cancer and cervix carcinoma. It is recommended that interruptions in treatment are avoided for such patients and that corrections are considered should a gap in treatment occur to minimise any prolongation of the overall treatment time.

Some publications discuss the 4 Rs of radiobiology, excluding radiosensitivity.

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Some publications discuss the 4 Rs of radiobiology, excluding radiosensitivity.

Related Articles: Alpha beta ratio, Fractionation, Interruption of treatment, Radiosensitivity, Repair, Repopulation, Redistribution, Reoxygenation

Therapeutic effect

(Radiotherapy) The effectiveness of a radiotherapy treatment is measured by its ability to obtain tumour control without inducing serious complications due to the irradiation of normal tissues.

In general, the relationship between dose and radiation effect, that is the dose–response curve, is sigmoid (S) in shape for both tumour control and normal tissue complications although the curve is often steeper for normal tissue damage than for tumour control. For the treatment to be effective, the response curve of normal tissues must lie to the right of the tumour response. Figure T.10 illustrates such a situation if the blue curves represent the tumour response and the red curves represent normal tissue.

The therapeutic index or therapeutic ratio is often used as a measure of the efficacy of treatment. It is the ratio of the tumour response for a fixed level of normal-tissue complication. In Figure T.10, the therapeutic ratio for the situation represented by the solid-line curves (Case A) is better than that for the dashed-line curves (Case B). In Case A, 95% probability of tumour control is possible for a 5% incidence of complications (the level generally regarded as acceptable in clinical radiotherapy for radical treatment) but in Case B this complication level only gives a 12% probability of tumour control. The therapeutic window is the horizontal distance between the normal tissue dose–response curve and the tumour control dose–response curve for a fixed level of normal-tissue complication. This defines the range of treatment doses that can be delivered without inducing an unacceptable level of normal tissue damage. In Figure T.10, the therapeutic window is the horizontal distance between the NTCP curve and TCP curve this is clearly larger for Case A than for Case B.

Both the solid and dashed curves may represent the same tumour and normal tissue combination but under different conditions, for example different fractionation regimes. Generally, the prolongation of treatment by the use of fractionation has been shown to be beneficial in many cases (for more details see the article on Fractionation). The use of a radiosensitiser or radioprotector may improve the therapeutic ratio if it produces a favourable differential effect (for more information see the article on Radiosensitisers).

Related Articles: Dose–response model, Fractionation, Normal tissue control probability, Radiosensitisers, Sigmoid dose–response curve, Tumour control probability

Therapeutic efficacy

(Radiotherapy) The therapeutic efficacy of a radiotherapy treatment is measured by its ability to obtain tumour control without inducing serious complications due to the irradiation of normal tissues. For more information, see the article on Therapeutic effect.

Related Articles: Adverse effect, Therapeutic effect, Tumour control

Thermal index (TI)

(Ultrasound) The thermal index is an indicator of the relative risk of thermal bio-effects resulting from diagnostic ultrasound.

Ultrasound may be converted to heat in tissue though absorption processes. Heating is dependent on the ultrasound power and intensity which in turn are dependent on the mode used – B-mode, colour flow, PW spectral Doppler and M-mode – and the scanner settings including power output, focal depth and frequency. The heating is also dependent on the tissue type and the proximity of the transducer face which can itself be a source of heat.

The thermal index is defined by

\[ TI = \frac{W_p}{W_{deg}} \]

where

- \( W_p \) is the power output
- \( W_{deg} \) is the estimated power necessary to raise the target tissue by 1°C

TI is calculated based on modelling of three different scanning conditions: scanning soft tissue which is described by TIS, scanning where there is bone in the focal region of the scan, TIB, and scanning where bone is close to the surface TIC, used in transcranial imaging, for example. Models have been produced to account for scanning (B-mode and colour flow) and nonscanning (M-mode and spectral Doppler) modes.

The output display standards (ODS) were put forward by the American Institute of Ultrasound in Medicine (AIUM) and the National Electrical Manufacturers Association (NEMA) in 1992 as a guide for users to monitor the output level, and by association, relative risk of ultrasound scanners. These have been adopted by the Food and Drug Administration (FDA). There is a requirement to display TIs greater than 1 on the system with the onus on the operator to keep within safe limits. No limits have been set by the FDA but several bodies have recommended upper limits for use in various clinical examinations.

Abbreviation: TI = Thermal index


Thermal neutrons

(General) A thermal neutron is a free neutron (one that is not bound within an atomic nucleus) that exists in thermal equilibrium with its surrounding material.

- Thermal neutrons are relatively slow and have a large cross-sectional area for fission interaction, making them desirable in certain nuclear chain reactions. They are also used in production of radio-isotopes through neutron activation of or neutron capture by the target material, typically in a nuclear reactor.

In radiotherapy, thermal neutrons are produced from repeated photon scatter down the bunker maze: polythene-loaded boron or lithium maze linings are often used to provide thermal neutron capture.

Related Article: Neutron capture cross section

Thermal stress

(Diagnostic Radiology) The reason for thermal stress of the anode of an x-ray tube is its rapid heating and cooling (during and after the x-ray exposure), which leads to rapid mechanical expansion and shrinking of the target material. This way after time the thermo-mechanical stress causes cracks on the tungsten surface (see Figure T.11). The cracks not only decrease the life of the x-ray tube (damaging the anode), but also uneven the target surface. The cracked anode surface causes scattering or absorbing (in the cracks) of some of the x-rays quanta, and thus decreases the tube efficiency.

In order to minimise this effect rhenium is added to the tungsten target (approximately 2%–10% Re/W metal alloy) and special care is taken for removing the heat accumulated in the anode. The x-ray tubes with rotating anode have all their front surface covered with this alloy, although only part of it is bombarded – the thermal path.

Another way to deal with the thermal stress is to manufacture the anode with radial slits, which allow the thermal expansion of the target without causing cracks (see the diagram in article Rotating anode).

Related Articles: Rotating anode, Target

Thermionic emission

(Diagnostic Radiology) Thermionic emission is emission of electrons or ions from the surface of a material, due to thermal energy greater than the electrostatic forces keeping these charges bound with the material. This effect is used in the x-ray tube cathode filament to produce thermal electrons. The density of the thermionic (thermal) emission current is described with the Richardson equation (called also Richardson–Dushman equation):

\[ J_0 = A_0 T^2 e^{-w/kT} \]

where

- \( J_0 \) is the density of the emission current
- \( T \) is the temperature of the emitter (in Kelvin)
- \( k \) and \( w \) are constants (\( k \), Boltzmann constant, \( w \), work function, for tungsten = 4.5 eV)
- \( A_0 \) is the constant depending of the material of the emitter (for tungsten = 60 A/cm²/K²)
**Related Articles:** Cathode, Filament heating, Filament current, Tube current


### Thermoluminescent dosimeter (TLD)

*(Radiation Protection)* Thermoluminescence is the physical phenomenon of emission of optical radiation in the form of prompt fluorescence when a material is heated. Many crystals, for example diamonds and chemical compounds such as lithium fluoride (LiF), can absorb ionising radiation and store it by moving electrons from the valence band to the conduction band and then capturing them at the trapping centres (metastable states) within the bandgap, below the bottom of the conduction band. Holes, created in the valence band, can move through the crystal and be trapped by the trapping centres situated above the top of the valence band (Figure T.12a). The probability that an electron escapes from the trap is proportional to the Boltzmann constant: \( \exp(-E/k_B T) \), where \( E \) is energy, \( k_B \) is Boltzmann’s constant and \( T \) is temperature. The life time of the trapped state can be large (up to hundreds of years) if \( E \) is large.

The trapped electrons can be detected through heating. When the temperature reaches the value determined by the energy level of the trap, the trapped electrons can move back to the conduction band and then recombine with the emitted visible radiation. At a temperature lower than that required to free the trapped holes, the electrons may migrate from their trap to a near trapped hole and recombine with the emitted visible photons (Figure T.12b). The amount of light released is proportional to the energy of ionising radiation absorbed by the thermoluminescent (TL) detector.

The equipment needed to perform thermoluminescent dosimetry (TLD) consists of a tray where the detector is placed and can be heated, a photomultiplier tube (PM) to measure the light emitted as the temperature is gradually increased, and associated electronics (Figure T.13). An oven to anneal the TLDs in between irradiations and restore their original sensitivity is also needed.

The emitted light as a function of the detector temperature may be graphically represented as a glow curve (Figure T.14). The radiation dose is proportional to the area under the glow curve. The TLDs can be used to measure absorbed doses in the range of \( 10^{-7} \) to \( 10^3 \) Gy.

The loss of trapped electrons at room temperature is called ‘fading’ and appears if the energy levels of the traps are very near to the edge of the band gap.

The choice of TLD detector should take into account the atomic number of the thermoluminescent material, the fading characteristics and the relation between the measured signal and the absorbed energy of radiation. TLDs based on LiF are mostly used because of their good energy response, due to the fact that the average atomic number of lithium fluoride is 8.2, very close to soft tissue, which has a value of 7.5. LiF may be doped with Cu, Mg, P and Ti.

TLDs are produced in the form of powder, discs (Figure T.15), chips, rings, rods and plates. LiF detectors enriched with Li-6 can be used to measure thermal (slow) neutron doses through the \((n, \alpha)\) reaction.

The thermoluminescent dosimeters work in passive mode but can be used many times because of their recyclability. However, they require individual or batch calibration against a standard. The TLDs are frequently used to monitor the dose of radiation workers and to measure patient doses undergoing radiological procedures.

**Abbreviation:** TLD = Thermoluminescent detector.

**Related Articles:** Dose, Radiation dosimetry, Radiation exposure


### Thermostat

*(General)* A thermostat is an on–off electrical temperature sensor which may be used in a thermal control process.
The thermostat is placed in the area where temperature is to be measured or clamped directly to the object being monitored. Both preset and adjustable thermostats act to sense temperature and compare it against a predefined value, switching an electrical circuit when that temperature is exceeded.

Typically the thermostat has a ‘hysteresis region’, such that once switched in one direction, it will not switch back until a predefined change in temperature has occurred. This prevents a race condition of ‘chatter’ of the electrical contacts when the threshold temperature is reached.

Thermostats are used as control sensors in both cooling and heating systems, and also on occasion as safety cut-outs. Thermostats only allow for on–off control, while more sophisticated control systems may use proportional sensors (see Temperature control).

Related Articles: Temperature control, Thermal probe, Temperature probe

**Thimble chamber**

(Radiotherapy) Gas-filled detectors represent probably the most widely used class of radiation detectors. The detection is based on the collection of ions produced in the sensitive volume of the detector as the radiation interacting in a gas-filled volume produces ion pairs in the volume through the process called ionisation. The essential parts of an ionisation chamber are two electrodes kept at different potentials. The electrode to which the measuring instrument is attached is called the collecting electrode. The collecting electrode is ordinarily kept at different potentials. The electrode to which the measurement is attached is called the collecting electrode. The collecting electrode is ordinarily but not necessarily at a potential close to ground potential while the other electrode, which is ordinarily kept at a constant voltage of several hundred volts, is called the high-voltage electrode. A number of commercially available ionisation chambers are generally used in radiotherapy dosimetry having air as a filling gas. An ionisation chamber is intended to measure the absorbed dose in a point in a medium and this requirement imposes an upper limit on the chamber dimension while its sensitivity and limitation in the mechanical design impose a lower limit. As the ionisation methods depend on the Bragg–Gray cavity theory and its extension by Spencer and Attix there is no explicit restriction on the geometrical shape of the cavity while the choice of the materials for wall, gas and electrodes as well the chamber wall thickness are governed by the cavity theory requirements. Generally the ionisation chambers are divided into three geometrical shapes: cylindrical or thimble, parallel plane and spherical. The shape of the ionisation chamber influences its spatial resolution when the spatial gradient of the radiation field is large. Figure T.16 shows the schematic diagram of a thimble chamber with the indication of the specific materials used for the chamber construction. However a variety of materials can be used requiring different correction factors.

The characteristics of some thimble ionisation chambers are given in Table T.1.

A thimble ionisation chamber type may be used for the calibration of beams of medium energy x-rays above 80 kV and an HVL of 2 mm aluminium, 60 Co gamma radiation, high energy photon beams, electron beams with energy above 10 MeV approximately, and therapeutic proton and heavy ion beams. Generally the calibration of all the high energy photon and electron beams are performed in a phantom as stated in the dosimetry protocols. In case of measurement in air the thimble chamber wall must be sufficient to provide electronic equilibrium and this condition is obtained adding a build-up cap to reach a wall thickness at least equal to \( d_{\text{min}} \) (Figure T.17a through c).

Depth dose measurements usually require thimble-type chambers with active volumes ranging from 0.1 to 1.0 cm\(^3\). Their walls are most often made from graphite or plastic (nylon, A-150 plastic). A variety of smaller chambers are available for measurement of field size factors with small fields. Miniature ionisation chambers with volumes less than 0.01 cm\(^3\) have also been developed.

Some typical thimble ionisation chambers are shown in Figures T.18 and T.19.

**Thimble ionisation chamber**

(Radiation Protection) See Thimble chamber

**Thin-film technology (TFT)**

(General) The advent of thin-film technology has allowed the development of all modern computing and therefore all digital medical imaging modalities. Thin films are fabricated by the deposition of individual atoms onto the substrate. A thin film is categorised as a low dimensional material produced by condensing atoms or molecules individually, with a final thickness of usually less than a few microns. Thin films are deposited onto a substrate by thermal evaporation, chemical decomposition (chemical vapour deposition CVD) and/or the evaporation of source material by irradiating it with energetic species or photons (sputtering). The properties of thin films are governed by the deposition process.

**Thin-layer chromatography (TLC)**

(Nuclear Medicine) Thin-layer chromatography, TLC, is a method used for determination of radiochemical purity and of various radiochemical impurities in radiopharmaceutical preparations, especially for \(^{99m}\)Tc-radiopharmaceuticals. This method is based on two phases, a stationary phase placed on glass plates or on aluminium or plastic foils and a mobile phase (solvent).
## TABLE T.1

### Characteristics of Some Thimble Ionisation Chambers

<table>
<thead>
<tr>
<th>Ionisation Chamber Type</th>
<th>Cavity Volume (cm³)</th>
<th>Cavity Length (mm)</th>
<th>Cavity Radius (mm)</th>
<th>Wall Material</th>
<th>Wall Thickness (g/cm²)</th>
<th>Central Electrode Material</th>
</tr>
</thead>
<tbody>
<tr>
<td>Capintec PR-05P mini</td>
<td>0.07</td>
<td>5.5</td>
<td>2.0</td>
<td>C-552</td>
<td>0.220</td>
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<td>11.5</td>
<td>2.0</td>
<td>C-552</td>
<td>0.220</td>
<td>C-552</td>
</tr>
<tr>
<td>Capintec PR-06C/G Farmer</td>
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<td>22.0</td>
<td>3.2</td>
<td>C-552</td>
<td>0.050</td>
<td>C-552</td>
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<td>3.2</td>
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<td>Capintec PR-06C/G Farmer</td>
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<td>3.2</td>
<td>C-552</td>
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<tr>
<td>Exradin A2 Spokas (2 mm cap)</td>
<td>0.53</td>
<td>11.4</td>
<td>4.8</td>
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<td>11.4</td>
<td>4.8</td>
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<td>NE 2505/A Farmer</td>
<td>0.6</td>
<td>24.0</td>
<td>3.0</td>
<td>Nylon 66</td>
<td>0.063</td>
<td>Aluminium</td>
</tr>
<tr>
<td>NE 2505/3, 3A Farmer</td>
<td>0.6</td>
<td>24.0</td>
<td>3.2</td>
<td>Graphite</td>
<td>0.065</td>
<td>Aluminium</td>
</tr>
<tr>
<td>NE 2505/3, 3B Farmer</td>
<td>0.6</td>
<td>24.0</td>
<td>3.2</td>
<td>Nylon 66</td>
<td>0.041</td>
<td>Aluminium</td>
</tr>
<tr>
<td>NE 2571 Farmer</td>
<td>0.6</td>
<td>24.0</td>
<td>3.2</td>
<td>Graphite</td>
<td>0.065</td>
<td>Aluminium</td>
</tr>
<tr>
<td>NE 2581 Farmer (PMMA cap)</td>
<td>0.6</td>
<td>24.0</td>
<td>3.2</td>
<td>A-150</td>
<td>0.041</td>
<td>A-150</td>
</tr>
<tr>
<td>NE 2581 Farmer (polystyrene cap)</td>
<td>0.06</td>
<td>24.0</td>
<td>3.2</td>
<td>A-150</td>
<td>0.041</td>
<td>A-150</td>
</tr>
<tr>
<td>NE 2561/2611 Sec. Std</td>
<td>0.33</td>
<td>9.2</td>
<td>3.7</td>
<td>Graphite</td>
<td>0.090</td>
<td>Aluminium (hollow)</td>
</tr>
<tr>
<td>PTW 23323 micro</td>
<td>0.1</td>
<td>12.0</td>
<td>1.6</td>
<td>PMMA</td>
<td>0.197</td>
<td>Aluminium</td>
</tr>
<tr>
<td>PTW 23331 rigid</td>
<td>1.0</td>
<td>22.0</td>
<td>4.0</td>
<td>PMMA</td>
<td>0.060</td>
<td>Aluminium</td>
</tr>
<tr>
<td>PTW 23332 rigid</td>
<td>0.3</td>
<td>18.0</td>
<td>2.5</td>
<td>PMMA</td>
<td>0.054</td>
<td>Aluminium</td>
</tr>
<tr>
<td>PTW 23333 (3 mm cap)</td>
<td>0.6</td>
<td>21.9</td>
<td>3.1</td>
<td>PMMA</td>
<td>0.059</td>
<td>Aluminium</td>
</tr>
<tr>
<td>PTW 23333 (4.6 mm cap)</td>
<td>0.6</td>
<td>21.9</td>
<td>3.1</td>
<td>PMMA</td>
<td>0.053</td>
<td>Aluminium</td>
</tr>
<tr>
<td>PTW 30001 Farmer</td>
<td>0.6</td>
<td>23.0</td>
<td>3.1</td>
<td>PMMA</td>
<td>0.045</td>
<td>Aluminium</td>
</tr>
<tr>
<td>PTW 30010 Farmer</td>
<td>0.6</td>
<td>23.0</td>
<td>3.1</td>
<td>PMMA</td>
<td>0.057</td>
<td>Aluminium</td>
</tr>
<tr>
<td>PTW 30002/30011 Farmer</td>
<td>0.6</td>
<td>23.0</td>
<td>3.1</td>
<td>Graphite</td>
<td>0.079</td>
<td>Graphite</td>
</tr>
<tr>
<td>PTW 30004/30012 Farmer</td>
<td>0.6</td>
<td>23.0</td>
<td>3.1</td>
<td>Graphite</td>
<td>0.079</td>
<td>Aluminium</td>
</tr>
<tr>
<td>PTW 30006/30013 Farmer</td>
<td>0.6</td>
<td>23.0</td>
<td>3.1</td>
<td>PMMA</td>
<td>0.057</td>
<td>Aluminium</td>
</tr>
<tr>
<td>PTW 31002 flexible</td>
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<td>6.5</td>
<td>2.8</td>
<td>PMMA</td>
<td>0.078</td>
<td>Aluminium</td>
</tr>
<tr>
<td>PTW 31003 flexible</td>
<td>0.3</td>
<td>16.3</td>
<td>2.8</td>
<td>PMMA</td>
<td>0.078</td>
<td>Aluminium</td>
</tr>
<tr>
<td>SNC 100730 Farmer</td>
<td>0.6</td>
<td>24.4</td>
<td>3.5</td>
<td>PMMA</td>
<td>0.060</td>
<td>Aluminium</td>
</tr>
<tr>
<td>SNC 100740 Farmer</td>
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<td>3.5</td>
<td>Graphite</td>
<td>0.085</td>
<td>Aluminium</td>
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<tr>
<td>Victoreen Radocon III 550</td>
<td>0.3</td>
<td>4.3</td>
<td>2.5</td>
<td>Delrin</td>
<td>0.529</td>
<td></td>
</tr>
<tr>
<td>Victoreen Radocon II 555</td>
<td>0.1</td>
<td>23.0</td>
<td>2.4</td>
<td>Polystyrene</td>
<td>0.117</td>
<td></td>
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<tr>
<td>Victoreen 30–348</td>
<td>0.3</td>
<td>18.0</td>
<td>2.5</td>
<td>PMMA</td>
<td>0.060</td>
<td></td>
</tr>
<tr>
<td>Victoreen 30–351</td>
<td>0.6</td>
<td>23.0</td>
<td>3.1</td>
<td>PMMA</td>
<td>0.060</td>
<td></td>
</tr>
<tr>
<td>Victoreen 30–349</td>
<td>1.0</td>
<td>22.0</td>
<td>4.0</td>
<td>PMMA</td>
<td>0.060</td>
<td></td>
</tr>
<tr>
<td>Victoreen 30–361</td>
<td>0.4</td>
<td>22.3</td>
<td>2.4</td>
<td>PMMA</td>
<td>0.144</td>
<td></td>
</tr>
<tr>
<td>Scdx-Wellhöfer IC 05</td>
<td>0.08</td>
<td>4.0</td>
<td>3.0</td>
<td>C-552</td>
<td>0.068</td>
<td>C-552</td>
</tr>
<tr>
<td>Scdx-Wellhöfer IC 06</td>
<td>0.08</td>
<td>4.0</td>
<td>3.0</td>
<td>C-552</td>
<td>0.068</td>
<td>C-552</td>
</tr>
</tbody>
</table>

*Capintec PR-05P mini* and *Capintec PR-05 mini* refer to the same type of chamber with different specifications.

*Farmer shortened* refers to chambers with modified dimensions or materials.

*Scdx-Wellhöfer IC 05* and *Scdx-Wellhöfer IC 06* are the same type with slight variations in specifications.

---

Thin-layer chromatography (TLC) 748 Thin-layer chromatography (TLC)
Thin-layer chromatography (TLC)

A small aliquot of the radiopharmaceutical is placed at least 1 cm from the end of the plate (application point). The plate is dipped in a covered tank containing the mobile phase, keeping the application point above the solution, and the solvent front is allowed to migrate a certain distance from the application point. Different components of the radiopharmaceutical will distribute themselves along the plate depending on the stationary phase, the solvent and on the solubility of the component. The application point and solvent front should carefully be marked to be able to calculate the \( R_f \) value. The plate is dried and the activity distribution is measured either by scanning the plate with a collimated NaI(Tl)-detector or by cutting the foil into small segments and measuring the activity with a NaI(Tl)-well counter. The distance each component migrates is characterised by the \( R_f \) (retention factor) value. For components migrating with the solvent front the \( R_f \) value is 1 and for the components remaining at the application point the \( R_f \) value is 0.

The stationary phase is available as, for example silica gel, reversed-phase silica and aluminium oxide. Most used are instant TLC (ITLC) plates made of fibreglass sheets impregnated with, for example silica gel. The advantage of ITLC is that it is rapid and accurate due to a fine mesh material that increases the migration properties. Whereas standard TLC takes more than 30 min to develop, ITLC may take less than 5 min.

### TABLE T.1 (continued)

**Characteristics of Some Thimble Ionisation Chambers**

<table>
<thead>
<tr>
<th>Ionisation Chamber Type</th>
<th>Cavity Volume (cm³)</th>
<th>Cavity Length (mm)</th>
<th>Cavity Radius (mm)</th>
<th>Wall Material</th>
<th>Wall Thickness (g/cm²)</th>
<th>Central Electrode Material</th>
</tr>
</thead>
<tbody>
<tr>
<td>Scdx-Wellhöfer IC 10</td>
<td>0.14</td>
<td>6.3</td>
<td>3.0</td>
<td>C-552</td>
<td>0.068</td>
<td>C-552</td>
</tr>
<tr>
<td>Scdx-Wellhöfer IC 15</td>
<td>0.13</td>
<td>5.8</td>
<td>3.0</td>
<td>C-552</td>
<td>0.068</td>
<td>C-552</td>
</tr>
<tr>
<td>Scdx-Wellhöfer IC 25</td>
<td>0.25</td>
<td>10.0</td>
<td>3.0</td>
<td>C-552</td>
<td>0.068</td>
<td>C-552</td>
</tr>
<tr>
<td>Scdx-Wellhöfer IC 28</td>
<td>0.3</td>
<td>9.0</td>
<td>3.1</td>
<td>C-552</td>
<td>0.068</td>
<td>C-552</td>
</tr>
<tr>
<td>Scdx-Wellhöfer IC 69 Farmer</td>
<td>0.67</td>
<td>23.0</td>
<td>3.1</td>
<td>Delrin</td>
<td>0.056</td>
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<tr>
<td>Scdx-Wellhöfer IC 70 Farmer</td>
<td>0.67</td>
<td>23.0</td>
<td>3.1</td>
<td>Graphite</td>
<td>0.068</td>
<td>Aluminium</td>
</tr>
</tbody>
</table>


**FIGURE T.17** Variation of the thimble chamber response with the wall thickness. (a) Thimble chamber; (b) build-up cap; (c) cap fitted the chamber.

**FIGURE T.18** 0.6 cm³ thimble chamber.

**FIGURE T.19** Microchamber.
Commonly used solvents are, for example 85% methanol, acetate, methyl ethyl ketone (MEK), acetone and 0.9% NaCl. The chromatographic separation of the components depends on the system and several stationary and mobile phases can be combined to not only obtain the radiochemical purity but also the amount of the various impurities in the preparation.

To avoid artefacts the TLC plates must be handled carefully and test conditions must be as similar as possible between measurements. It is important that the solid phase must be dry and that the sample spot is small and should dry before being placed in the solvent tank. The case of prolonged air contact, however, may cause oxidation of the $^{99m}$Tc-chelate and technetium components may also bind to the solid phase.

**Related Articles:** Gel chromatography, R$_{f}$-value


### Thiosulphate in film processing

**(Diagnostic Radiology)** Compounds of thiosulphate ($S_{2}O_{3}^{2-}$), such as sodium thiosulphate (also known as hypo) or ammonium thiosulphate, are used as the fixer in the film processing cycle. After a film is removed from the developer it is then passed through the fixing solution. The fixer performs several actions including neutralizing and stopping the development process, removing the undeveloped silver halide grains from the film, and hardening the emulsion.

### Thomson scattering

**(Radiation Protection)** Thomson scattering named after J.J. Thomson is the scattering of electromagnetic radiation by a charged particle, mainly electrons. When an electromagnetic wave passes near an electron, the electron is momentarily accelerated by the electric field of the wave and, as a consequence, it radiates energy. The cross section of this process, called the Thomson classical scattering coefficient, may be derived from classical physics and has the same value for all photon energies. In this type of scattering, the incident photon interacts with a free electron but the interaction is elastic and the electron is given no energy; it is merely scattered from the beam.

**Related Article:** Elastic scattering

### Thorium

**(General)**

<table>
<thead>
<tr>
<th>Property</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Symbol</td>
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<tr>
<td>Element category</td>
<td>Transition metal</td>
</tr>
<tr>
<td>Mass number A</td>
<td>232</td>
</tr>
<tr>
<td>Atomic number Z</td>
<td>90</td>
</tr>
<tr>
<td>Atomic weight</td>
<td>232.04 g/mol</td>
</tr>
<tr>
<td>Electronic configuration</td>
<td>1s$^{2}$ 2s$^{2}$ 2p$^{6}$ 3s$^{2}$ 3p$^{6}$ 3d$^{10}$ 4s$^{2}$ 4p$^{4}$ 4d$^{10}$ 4f$^{14}$ 5s$^{2}$ 5p$^{6}$ 5d$^{10}$ 6s$^{2}$ 6p$^{6}$ 6d$^{2}$ 7s$^{2}$</td>
</tr>
<tr>
<td>Melting point</td>
<td>2023 K</td>
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<tr>
<td>Boiling point</td>
<td>5061 K</td>
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<tr>
<td>Density near room temperature</td>
<td>11.72 g/cm$^3$</td>
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</table>

**History:** Thorium was discovered in 1828 by Berzelius from a sample of a previously unknown ore now known as thorite (ThSiO$_4$). It is found in small amounts in most rocks and soils, but is usually commercially obtained from monazite along with other rare-earth metals. The element has limited applications as an alloying addition to magnesium and to tungsten. It has potential uses as a fertile material in nuclear fuels. There are several uses for the very high melting point thorium oxide (ThO$_2$), including gas mantles, high temperature crucibles and high refractive index glass for lenses.

**Medical Applications:** Disused contrast agent – Thorium oxide was used as Thorotrast (a contrast agent) in x-ray diagnostics. This was discontinued due to its carcinogenic nature.

**Related Article:** Contrast agent

### Three-dimensional ultrasound imaging (3D imaging)

**(Ultrasound)** Three-dimensional (3D) ultrasound imaging is the acquisition and display of ultrasound from a volume of tissue. Conventional B-mode and colour flow imaging obtains data from a plane of thin slice thickness, essentially a 2D image. The operator moves the transducer to obtain a series of real-time images from different positions in the elevation plane. In practice free movement of the transducer by rotation and translation are used to build up an overall picture of the tissue under investigation.

In 3D imaging the acquisition of echoes from a tissue volume is used to provide a volume of data which can be viewed and analysed at the time of the scan of after data acquisition. The terms 3D and 4D are used to describe volume imaging of ultrasound. In general, 3D encompasses static and dynamic 3D acquisition and display; 4D refers to specifically dynamic 3D imaging.

3D imaging can be thought of as having three distinct stages:

- Data acquisition
- Image processing and segmentation
- Presentation/display

There are currently three different methods for 3D acquisition which are commercially used (Figure T.20):

- Freehand movement in the elevation plane. This provides a 3D volume set but without spatial measurement of movement and distance in the elevation plane.
‘Wobbling’ or mechanically rotated arrays are usually curvilinear arrays which are swept in the elevation plane. This technique can be used at frame rates of a few Hz to provide moving images of, most commonly foetal movement. Distances in the elevation plane are known from the angle at which individual images are obtained.

2D arrays insonate a volume of tissue from plane surface. Frame rates of >15Hz are possible and the technique is also used for 3D colour flow imaging. The location of all points in the axial, lateral and elevation planes are established allowing measurement of linear, area and volume dimensions.

Other 3D techniques include external location of transducer position including optical and magnetic methods.

Processing can be real-time or following data capture, including off-line processing. Segmentation can be difficult for ultrasound since echo levels from surfaces are dependent on the beam/interface angle. The most common surface rendering images are of the heart and foetus where there is a strong fluid/tissue interface.

The display of 3D ultrasound can be of a surface (Figure T.21) or by examining different planes within a block of data (3). The latter allows imaging in the C-plane, which has been shown to be beneficial in coronal imaging of breasts. In surface and planar imaging, the operator has a range of tools to alter the planes imaged and the post-processing of images. Surface rendered images can similarly be viewed from different angles and with different rendering parameters (Figure T.22).

Related Article: Matrix arrays

Three-phase AC
(Diagnostic Radiology) A continuous series of three overlapping AC cycles offset by 120°. Three-phase power is used for all large scale electrical distribution systems.

Three-phase generator
(Diagnostic Radiology) Power x-ray equipment with classical high voltage generator use three-phase mains supply. This ensures sufficient power and produces kV with less pulsation (this way reducing both – the absorbed dose and image noise). The high voltage transformer of these generators requires different types of secondary winding connections.

FIGURE T.20 Three different types of transducer formats for data acquisition. (a) Conventional array moved manually in the elevation plane. (b) Mechanically rotated curvilinear array. (c) 2D matrix array.

FIGURE T.21 Surface views of 3D volume set of an aorta.
The three phase generator producing kV with six-pulse waveform uses high voltage transformer with secondary winding, which is normally with star type connection (Figure T.23). The rectifiers in this case are usually in bridge-type connection. There are two types of electrical circuits – asymmetric (3 secondary windings and 6 diodes) and symmetric (6 secondary windings and 12 diodes). The six-pulse waveform ensures kV pulsations of the order of 14%. This is far less than the 100% pulsations of the 2 pulse generator using single-phase mains supply.

Further reduction of kV pulsations is achieved by the 12-pulse classical high voltage generator. In this case the secondary winding of the high voltage transformer is normally with star-delta connection (Figure T.24). This ensures kV pulsations of the order of 3%–4%. The electrical circuits of these generators (transformer and rectifiers) are also two types – asymmetric (6 secondary windings and 12 diodes) and symmetric (12 secondary windings and 12 diodes).

**Related Articles:** High voltage generator, High frequency generator, High voltage circuit, voltage waveform

**Further Reading:** Karadimov, D. 1978. Roentgen equipment, Technika, Sofia, Bulgaria.

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**Three-phase rectifier**

*Diagnostic Radiology* The type of rectifying circuitry of an x-ray three-phase generator (classical type) – see typical circuits (usually diodes with bridge-type connection) at the eponymous article.

**Related Articles:** High voltage generator, Three-phase generator, High voltage circuit

**Three-phase transformer**

*Diagnostic Radiology* Powerful equipment (e.g. power x-ray generators) use three-phase power supply as opposed to the normal mains 220 V supply. The transformers used with three-phase mains are normally with three primary coils (for each phase) and a number of secondary coils. The most often used connection of the secondary coils is star, delta or star-delta. For illustration of these connections see the diagrams in the article *Three-phase generator*.

**Threshold contrast detail detectability (TCDD)**

*Diagnostic Radiology* Threshold contrast detail detectability (TCDD) is a characteristic used to describe physical image quality by assessing the visibility of low contrast details. The use of TCDD tests in conjunction with DQE analysis has proved a practical and effective method for evaluating the performance of the whole digital x-ray imaging system.
The results of a TCDD test can be presented as a contrast detail diagram, or more conveniently as a threshold detection index \( H_f(A) \), defined by the following formula:

\[
H_f(A) = \frac{1}{(C_f(A) \times A^{0.5})}
\]

where \( C_f(A) \) is the threshold contrast corresponding to the last visible image detail of area \( A \). Threshold contrast \( C_f(A) \) is determined by the image taken of the contrast-detail test phantom, for example by CDRAD (Artinis Medical Systems, the Netherlands) or TO12 or TO20 (Leeds Test Objects Ltd, United Kingdom). An accepted protocol for the use of these test phantoms in routine testing has been in widespread use and is recommended in IPEM Report 32 (2010). The threshold detection index is usually plotted against the square root of detail area on double logarithmic axes (Figure T.25).

The observer views the image under a set of standard conditions, and counts the number of details visible in each row. The viewing results are then compared with tabulated data, relating the number of details seen to threshold contrast, for given x-ray beam quality. Alternatively, the contrast of the image details can be calculated for various thicknesses of the PMMA using the x-ray spectral and attenuation data derived from IPEM Report 78. Also software is available for automated scoring of CDRAD – this is more consistent and removes the subjectivity of the human observer from the evaluation process.

TCDD scores are dependent on detector air kerma (DAK) and therefore it is important to choose such a receptor dose for testing that allows a valid comparison with the reference data. For example, a DAK of 4 \( \mu \)Gy is assumed for direct comparison with reference data. For example, a DAK of 4 \( \mu \)Gy is assumed for direct comparison with reference data. For example, a DAK of 4 \( \mu \)Gy is assumed for direct comparison with reference data.

At commissioning, the TCDD results are compared to those from other similar systems, if available, and are used to set a baseline (reference TCDD curve) for future quality control tests.

Figure T.25 presents threshold detection index curve measured by using test phantoms CDRAD and TO12.

Alternatively, a single image quality factor can be used to track changes in image quality over a range of all detail sizes (Cowen and Workman 1992; Gallacher et al. 2003). This factor shows less variability than the scores for individual detail sizes. The image quality factor (IQF) is calculated by using the following formula:

\[
IQF = \frac{1}{n} \sum_{i=1}^{n} H_f(A_i) \left( \frac{K_{ref}}{K} \right)^{0.5}
\]

where

- \( H_f(A_i) \) is the threshold contrast detail index value for the image of interest
- \( H_f^0(A_i) \) is the threshold contrast detail index value for the reference image
- \( K \) is the DAK at the image plate
- \( K_{ref} \) is the DAK at the image plate for the reference image
- \( n \) is the number of different detail diameters visible in the image

**Related Articles:** Contrast detail (C-D) studies, Contrast detail


**Hyperlinks:** IPEM, KCARE

**Threshold detection**

*(General)* A threshold value refers to a particular level of a parameter (usually a voltage) which has been defined so as to discriminate between signals above and below that value. The result is therefore a binary logical one (above threshold versus below).

In ionising radiation detectors, the output signal usually consists of many short pulses of varying height each related to the incident energy of individual particles or rays. It is therefore possible to set a threshold voltage related directly to a specific particle or ray’s energy, the threshold energy, and only count those signal events which exceed this threshold. In this way it is possible to make a selective detector only counting particles above a certain energy.

It is also possible to specify multiple threshold energies or values and, by combining the outputs of a parallel group of threshold detectors with some basic logic, provide information in the form of count number against energy band or pulse height – a pulse height discriminator.

**Related Article:** Pulse height discriminator

**Threshold energy**

*(Diagnostic Radiology)* See Threshold detection

**Related Articles:** Pulse height discriminator, Threshold energy

**Threshold value**

*(Diagnostic Radiology)* See Threshold detection

**Related Articles:** Pulse height discriminator, Threshold detection
Thulium
(General)

<table>
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<th>Symbol</th>
<th>Tm</th>
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<td>Element category</td>
<td>Lanthanoid metal</td>
</tr>
<tr>
<td>Mass number A</td>
<td>169</td>
</tr>
<tr>
<td>Atomic number Z</td>
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<td>Melting point</td>
<td>1818 K</td>
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<td>Boiling point</td>
<td>2223 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>9320 kg/m³</td>
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</table>

**History:** Thulium was discovered in 1879 by Per Theodor Cleve when he extracted holmium and thulium oxides from impurities in erbium oxide. It is now obtained along with many other rare-earth elements by chemical separation using an ion exchange process applied to monazite sand ((Ce, La, Th, Nd, Y)PO₄). It is the least abundant of the naturally occurring rare-earth elements and has no significant commercial uses.

**Medical Applications:** Radiation source – Radioactive isotopes of thulium (such as thulium-171) have been used as brachytherapy seed sources. Another isotope, thulium-169, has a potential use as a radiation source for portable x-ray machines.

Lasers – Thulium lasers are often high power and well suited for use in medicine.

**Related Articles:** X-ray, Brachytherapy, Brachytherapy sources

**Thyratron**
(Diagnostic Radiology) Thyratron is a gas filled electronic tube, mainly used as high voltage switching device or controlled rectifier. The thyratron includes a cathode (more often hot filament cathode), control grid(s) and anode. The gas type varies and often includes xenon, neon and hydrogen. The electrical current through the device is controlled by varying the potential of the control grid. Thyratron is specially useful for fast commutation of high voltage. One of its many uses is for kV rectifiers and stabilisers (see the article about x-ray pulse-less generator). According to the number and type of the control grids the thyratron can be triode, tetrode or pentode. The thyratron was one of the most commonly used switches/rectifiers in x-ray classical high voltage generators, but is now replaced by semiconductor elements as thyristor and triac. However it is still very good for fast kV commutation due to its very short switching time.

**Related Article:** Pulse-less generator

**Thyroid radioiodine uptake measurements**
(Nuclear Medicine) The determination of the uptake of radioactive iodine by the thyroid gland was the first radioisotope test that required in vivo measurements. A certain amount of radioactive iodine, usually ¹²³I or ¹³¹I, is administered to the patient. The test consists of the determination of the fraction of the activity present in the thyroid at a later time.

The test has to consider the following parameters: (1) the amount of radioactivity present in the thyroid has to be determined accurately, thus the detector has to be calibrated and the thyroid position and volume determined to account for the tissue attenuation and self attenuation and (2) at the time of measurement only a fraction of the iodine is present in the thyroid and it is important to correct for the background signal (meaning the activity in the extrathyroidal tissue in the detectors field of view).

The instruments used are either single collimated probes or dedicated scintillation cameras. The measurements have to be calibrated with a standard activity and a proper phantom (neck phantom).

**Related Articles:** Iodine, Thyroid, Probe


**TI (inversion time)**
(Magnetic Resonance) See Inversion time (TI)

**Tilting table**
(Diagnostic Radiology) A table, typically for fluoroscopy, that can be tilted to raise the head of the patient during some procedures. In some countries the tilting table position with patient head lower than his feet is called Trendelenburg position.

**Time activity curve**
(Nuclear Medicine) When calculating the radiation dose to a specific organ or a patient one of the important parameters is the kinetics of the radio-compound in the patient (i.e. delivery, uptake (accumulation), metabolism, clearance in the specific organ). After injection different organs are studied to determine the spatial and temporal distribution of the administered activity. The spatial and temporal distribution for an individual compartment is called the time activity curve.

In the MIRD formalism the time activity curve is used to determine the cumulated activity which is an integration over the time activity curve from injection (t = 0) to total radionuclide clearance. Thus the cumulated activity is the total number of disintegrations in the organ from initial uptake to total clearance.

**Related Articles:** MIRD formalism, Cumulated activity


**Time average intensity**
(Ultrasound) Time average intensity is the intensity measured over a given cross-sectional area, averaged over time. The time average is taken over an integral number of pulse or field repetition periods. It will average out the variation in intensity resulting from the pulse-space ratio in pulsed ultrasound beams and give the average intensity at a field point ionsonated by multiple beams when in a scanning mode.

**Related Articles:** Intensity, Spatial average intensity, Pulse average intensity

**Time constant**
(General) A time constant is a measure of how quickly a system can respond to a step change in input parameter.

A time constant is usually used to describe the response a system which is exponential in form, either when it is slow to change (a), or transient (b) (Figure T.26).

The two forms have the mathematical expressions shown, and in both cases they are fully defined by the time constant τ, the time it takes each circuit to reach 63% of its final value in response to a step input change.

In the simple electronic circuits given in the following, their responses can be described using the time constant, and additionally can be shown to be related simply to the values of the circuit components shown (Figure T.27):

Their frequency responses are also related directly to their time constants, the low-pass filters cutting off signals higher than 1/2πτ Hz, while the high-pass filters pass signals higher than this frequency.
Time delay circuit

A time delay circuit is designed to respond to a change in input with a predefined delay. Both input and output are usually digital (on/off) and sometimes incorporate a relay in the output circuit to provide an electrically isolated but powerful switch.

Time delay circuits can be complete modular units, usually with a screw adjusted time delay given in seconds, and with specified input and output properties.

Time delay circuits can also be incorporated within more complex circuits, and vary in design depending on the required accuracy of the delay. The simplest type, often used in ‘power on reset’ circuits, uses the slow rise of voltage when a capacitor is charged through a resistor attached to the circuit’s DC power input line. This is fed to a comparator whose output changes when the input voltage rises above a set reference level (Figure T.28).

More advanced circuits often use the ubiquitous ‘555 timer’ integrated circuit which incorporates all the necessary active components and requires only the odd resistor and capacitor to produce a reliable and programmable delay.

Related Article: Time constant


Time delay integration

(Diagnostic Radiology) Time delay integration (TDI) is used in scanned-beam acquisition imaging systems (e.g. digital mammography). The detector of these systems is a multilinear array with charge coupled devices (CCDs). The image is acquired by scanning the object with a narrow fan beam of x-rays.

CCDs develop a charge on their individual sensors which is proportional to both radiation intensity and time of exposure. Thus they can integrate the effects of exposure over a period of time. In practice, this process may be taken further as the CCD device has a method of readout which transfers the charge on individual sensors in a linear fashion across the detector area, serially outputting the results for each line of sensors as the charge packets reach the edge.

In certain imagers with TDI the two processes are combined – the detector and x-ray source jointly being moved linearly in one direction while the charge packets move in the other direction. In this way, by clocking the charges across the sensors, whilst moving the detector at equivalent speed in the opposing direction, the radiation from one fixed part of the patient continues to fall on and integrate with the rest of the charge being detected from that part of the patient.

The TDI mode of operation of these systems improves sensitivity and reduces the influence of scattered radiation, but it takes longer time. The spatial resolution of these systems depends on the accurate alignment of the signal gathering CCD columns of the detector.

Time distance shielding (TDS rules)

(Radiation Protection) There are three means to protect people against external radiation exposure: time, distance and shielding. That the time of permanence in the radiation field will proportionally reduce the exposure of the irradiated individual can be grasped intuitively. To a certain extent, the same applies to the distance. In this case, however, there is no linear relationship, but an inverse square relation between the intensity of the radiation and the distance of the exposed individual from the source. This means that when the distance from the radiation source is doubled, there is one fourth of the exposure. Protection
can also be achieved by shielding the radiation source with appropriate materials (see Lead content). The attenuation of the radiation by this means follows an exponential relationship with the thickness of the shielding material.

Hyperlink: http://IAEA.org

Time gain compensation
(Ultrasound) See Depth gain compensation

Time interval difference imaging
(Nuclear Medicine) A difference image is the result of the pixel by pixel subtraction of one image from another. In time interval difference imaging, the two images are performed at different times.

In nuclear medicine this can be used to look at the washout of a tracer from a particular organ. An example of this is parathyroid imaging using Tc-99m MIBI. Over a period of time the tracer washes out of the thyroid revealing the parathyroid gland. It can also be used to look at the uptake in a tumour pre and post-therapy after registration and normalisation of the two images.

Time of flight
(Nuclear Medicine) Time of flight is a technique used in PET to determine the position of positron annihilation. Ideally, the time between two detected events can provide information about the localisation of the event along the line of response (LOR). If the annihilation event is off centre one of the two photons will be detected earlier, less than a nanosecond, than the other photon. The principal limiting factor today is that the rise time of light output in the common scintillators are to slow (1 cm depth resolution requires 66 ps timing resolution). The position along the LOR, \( \Delta d \), is given by

\[
\Delta d = \frac{\Delta t \times c}{2}
\]

where

\( \Delta t \) is the time difference between the two events
\( c \) is the speed of light (\( 3 \times 10^8 \text{ m/s} \))

Abbreviations: LOR = Line of response and PET = Positron emission tomography.

Related Article: PET

Time-of-flight (TOF)
(Magnetic Resonance) Time-of-flight (TOF) is MR angiography method that utilises the inflow effect (see Inflow effect).

In order to maximise signal enhancement from inflowing liquid and to reduce the signal from stationary tissue, a gradient echo (GRE) pulse sequence with a short TR is used. The strength of the signal is dependent on a number of parameters like slice thickness, flow velocity, flip angle, \( T_1 \) relaxation time and TR. Normally, in order to avoid pulsation flow artefacts gradients are also designed to refocus spins with a constant velocity (flow compensation).

Very often multiple two-dimensional (2D) inflow MRA, that is a number of thin contiguous 2D slices, are acquired with a fast gradient echo sequence. Subsequently, the 2D data are collected in the computer and a three-dimensional (3D) data set is formed and post-processed. In 3D inflow MRA, a thick slab or a volume is excited and the spatial encoding is performed in all three dimensions.

The major difference between the 2D and the 3D inflow MRA methods lies in their difference in slab thickness, causing different signal behaviour due to progressive saturation of the inflowing spins as they move deeper into the imaging slab. This phenomenon, which introduces a dependence upon flow direction, makes 2D inflow MRA more advantageous for low-velocity motion, since in 3D MRA, in this case a high signal is obtained at the entrance side while at the exit side the signal is low. On the other hand, the 3D technique gives better spatial resolution.

The most common method for the calculation of MR angiograms is the so-called ray-tracing technique. Using ray-tracing, a number of slices obtained in consecutive acquisitions of 2D images or a set of slices obtained in a 3D acquisition are stored in a data set. By following rays through the data set the maximum intensity is selected along that ray or projection (MIP). This value is taken as the pixel value for that pixel hit by the ray in the projected image plane.

By rotating the projection angle it is possible to obtain a 3D visualisation in any direction. It is also possible to exclude parts of the 3D data set and to perform the computation of the projection images through a selected subvolume of special interest. Vessels of minor interest can then be excluded and an enhanced image of interest may be obtained. A corresponding technique, more suitable for the outflow effect, is ray-tracing using a minimum intensity projection (Figure T.29).

Related Articles: Inflow effect, Maximum (minimum) intensity projection

Time of repetition
(Magnetic Resonance) See Repetition time (TR)

Time-of-flight techniques in PET
(Nuclear Medicine) See Time of flight

Timer
(Nuclear Medicine) A timer is used to set a time interval when a measurement is to be conducted. In radiology for x-ray exposure the timing device functions as an automatic exposure timer and a switch to control the current to the high-tension transformer and filament transformer. The face of the timer is calibrated in seconds and fractions of seconds. The timer controls the total time that the current passes through the x-ray tube and thus the time during which the roentgen rays are emitted. In nuclear medicine a timer is used to measure the number of registered pulses (proportional to impinging particles of photons on a detector) during a certain time interval. Then also the count rate is easily determined.

Tin
(General)

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Sn</th>
</tr>
</thead>
<tbody>
<tr>
<td>Element category</td>
<td>Metals</td>
</tr>
<tr>
<td>Mass number ( A )</td>
<td>120</td>
</tr>
<tr>
<td>Atomic number ( Z )</td>
<td>50</td>
</tr>
<tr>
<td>Atomic weight</td>
<td>118.710 kg/kg-atom</td>
</tr>
<tr>
<td>Electronic configuration</td>
<td>( 1s^2 \ 2s^2 \ 2p^6 \ 3s^2 \ 3p^6 \ 3d^{10} \ 4s^2 \ 4p^6 \ 4d^{10} \ 5s^2 \ 5p^3 )</td>
</tr>
<tr>
<td>Melting point</td>
<td>505.1 K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>2875 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>7310 kg/m³</td>
</tr>
</tbody>
</table>
Tissue tin (aka Stannum) is a ductile and malleable silvery-white metal with a highly crystalline structure. The major source of tin is the mineral cassiterite, in which tin is present as tin oxide, $\text{SnO}_2$. Tin is used in alloys, such as bronze, and as a coating to prevent corrosion of other metals.

**Medical Applications:** In nuclear medicine, tin (II) fluoride (stannous agent) is used to bind $^{99m}\text{TcO}_4^-$ (pertechnetate) to red blood cells to allow multi-gated acquisition (MUGA) imaging of the heart. The injected stannous agent binds with the patient’s red blood cells in vivo; this then acts as a reducing agent on the injected technetium causing it to bind to the red blood cells. A further use of tin in nuclear medicine is the imaging of sentinel nodes using a $^{99m}\text{Tc}$-tin colloid radiopharmaceutical.

**Related Articles:** Nuclear medicine imaging, Multi-gated acquisition, Pertechnetate, Tc-99m-Tin colloids

**Tissue**

*(General) CT number:* ~ 0 HU (water), varies from −120 HU (fat) to >400 HU (bone)

Biological tissue is a cellular organisational level between cells and organs. It is defined as a group of cells of the same origin, which provide a certain function. The functional grouping of several tissues form organs. The study of tissue is known as histology, or histopathology when in reference to disease. Tissues are studied by optical and electron microscopy as well as immunofluorescence.

Animal tissues are categorised into four types based on their morphology, including connective, epithelial, muscle and nervous tissues. Connective tissue consists of cells separated by an extracellular matrix that joins other tissues together and is flexible. Bone (osseous tissue) and blood are examples of connective tissues. Epithelial tissues are layers of cells that cover organ surfaces, such as the skin and the inner lining of the digestive tract. The layer protects the underlying organ by providing a barrier to the external environment due to its semi-permeability. The epithelium also allows secretion and absorption. Muscle tissue is an active contractile tissue producing force to enable motion, either externally or internally. Muscle tissue is further subdivided into visceral/smooth muscle, in the inner linings of organs; skeletal muscle, attached to bone for movement; and the cardiac muscle of the heart. The central and peripheral nervous systems are comprised of nervous tissue, including the brain, spinal cord, nerves and motor neurons.

**Related Articles:** Bone soft tissue interface, CT Number, Dose limiting tissue, Equivalent tissue air ratio (ETAR), Fat, Late response of normal tissue, Normal tissue complication probability (NTCP), Normal tissue dose, Normal tissue dose–response, Normal tissue reaction, Normal tissue toxicity, Tissue air ratio (TAR), Tissue compensation, Tissue contrast, Tissue deficit, Tissue equivalent material, Tissue heterogeneity, Tissue maximum ratio (TMR), Tissue phantom ratio (TPR), Tissue substitute, Tissue weighting factor, Water

**Tissue air ratio (TAR)**

*(Radiotherapy)* The tissue air ratio was introduced for the dosimetric calculations of firstly rotational radiotherapy, and more latterly, isocentric treatments, in which the SAD remains constant with the SSD varying with the patient contour.

It is the ratio of $D_Q$, the absorbed dose in tissue at point $Q$ on the central axis in a patient or phantom, to $D_Q^\text{air}$ the absorbed dose to a small mass of water in air at the same point $Q$ on the central axis. The set-up is illustrated in Figure T.30. It is defined as follows:

$$\text{TAR}(d,A,E) = \frac{D_Q(d,A,E)}{D_Q^\text{air}(air,A,E)}$$

**FIGURE T.29**  (a) Illustration of one slab from a 3D TOF data set (bright areas are vessels with prominent inflow effect). (b) Coronary MIP obtained from a 2D TOF sequence showing intracranial vessels. (Courtesy of Elna-Marie Larsson, Aalborg, Denmark.)

**FIGURE T.30** Set up for measuring the tissue air ratio (TAR). The field size $A$ is defined at point $Q$, which is normally placed at the isocentre.
In ICRU 24 the definition of TAR was given as follows:

The absorbed dose at a given point in a phantom

The absorbed dose which would be measured at the same point in free air within a volume of the phantom material just large enough to provide electronic equilibrium at the point of reference.

The denominator can also be thought of as the absorbed dose due to primary radiation only, and is normally measured with ionisation chambers and build-up caps. However there are difficulties with this definition when it comes to measurement:

- At high energies, the build-up volume needed to completely avoid scattered contribution is very large.
- Small field sizes below limit of build-up material diameter.
- Readings must be free of scattered radiation from the walls or floor.
- The actual measurement set-up is cumbersome, compared to PDD measurements which can be automated more easily.

In practice, TARs are often derived from measured PDD curves, as they can be related from first principles. The conversion process includes a Mayneord factor that is an inverse square law correction.

\[
\text{At } d = d_{\text{max}}, \text{ the TAR becomes identical to the peak scatter factor (PSF).}
\]

The TAR decreases as \( d \) increases further from \( d_{\text{max}} \). For constant depth and energy, TAR increases with increasing field size. For constant depth and field size, TAR increases with energy.

**Related Articles:** Peak scatter factor (PSF), Percentage depth dose (PDD), Scatter air ratio (SAR)

**Tissue compensation**

(Radiotherapy) To compensate for missing tissue or a sloping surface, a custom made bolus arrangement can be made that conforms to the patient’s skin on one side and yields a flat perpendicular incidence to the beam (see Figure T.31a). The result is an isodose distribution that is identical to that produced on a flat phantom; however, skin sparing is not maintained. This can be overcome by retracting the bolus (taking divergence into account) as in Figure T.31b.

A common material used for this kind of bolus is wax, which is essentially tissue equivalent and when heated is malleable and can be fitted precisely to the patient’s contour.

**Related Articles:** Bolus

**Tissue contrast**

(Diagnostic Radiology) See Chest radiography

**Tissue deficit**

(Radiotherapy) Because of the variation in body shape, the skin can present a surface that is not flat with respect to the radiation beam. At times there can be more or less tissue in the way of the beam and underlying treatment region (see Figure T.32). This will affect the dose distribution and the dose at the point of interest in Figure T.32. The situation can be regarded as one where there is a tissue deficit (i.e. missing tissue) and if the dose needs to be altered at the point of interest then some tissue equivalent material, bolus or a retractable compensator may have to be placed in the way of the beam. If bolus is used at the skin surface, the reduction in skin sparing will need to be taken into account.

**Tissue equivalent material**

(Radiotherapy) A material that responds in the same way as tissue does when irradiated is said to be tissue equivalent. For this to be the case the material has to have:

1. An effective atomic number close to tissue so that the photo-electric and pair production processes are contributing to absorption in the same way as tissue.
2. An electron density close to that of tissue so that Compton scatter and absorption is similar to that in tissue.
3. A density close to tissue so that any spatial measurements made in the phantom will be relevant to those made in tissue.

**Tissue heterogeneity**

(Radiotherapy) See Heterogeneity

**Related Articles:** Inhomogeneity correction factor, Heterogeneity

**Tissue maximum ratio (TMR)**

(Radiotherapy) The tissue maximum ratio (TMR) is a special case of tissue phantom ratio (TPR) where the reference depth is chosen to be the depth of dose maximum \( d_{\text{ref}} = d_{\text{max}} \). See Tissue phantom ratio (TPR) for details of the set-up.

The TMR ranges between 0 and 1, where it equals 1 for \( d = d_{\text{max}} \). For constant field size and energy, the TMR decreases with...
increasing depth. The TMR increases with both increasing field size and increasing energy, at constant depth.

**Related Article:** Tissue phantom ratio (TPR)

### Tissue phantom ratio (TPR)

*(Radiotherapy)* The tissue phantom ratio was introduced to cope with the difficulties of measuring the tissue air ratio (TAR) in MV set-ups, where the build-up required for the ‘in-air’ measurement becomes impractical. The TPR is defined as the ratio of the dose in a phantom at point $Q$ on the central axis, to the dose at the same point $Q$ in a phantom at a reference depth $d_{ref}$. It is equivalent to the ratio of the corresponding dose rates (Figure T.33):

$$TPR(d,A,E) = \frac{D_Q}{D_{Q_{ref}}} = \frac{\bar{D}_Q}{\bar{D}_{Q_{ref}}}$$

The measurement set-up is shown in Figure T.33. It is measured in water or water substitute solid phantoms by keeping the detector at a constant distance from the source and varying the depth of material.

When $d_{ref}$ is chosen to be $d_{max}$, the TPR becomes the tissue maximum ratio (TMR).

**Related Articles:** Peak scatter factor (PSF), Percentage depth dose (PDD), Scatter air ratio (SAR), Tissue air ratio (TAR)

### Tissue substitute

*(Radiotherapy)* A tissue substitute is a material that can be used as tissue equivalent in either in a phantom to make dosimetric measurements or to provide necessary build-up material on the skin while the patient is being treated. Materials such as wax and moist gauze are often used. Barley and tissue equivalent gels, which are commercially available, are also used. See also Tissue equivalent material.

### Tissue weighting factor

*(Radiotherapy)* Tissue weighting factors are similar to the radiation weighting factors applied to convert absorbed dose to equivalent dose. They are assigned to different body tissues and are used to convert equivalent dose to effective dose using Equation T.1. The value of each weighting factor is based upon the radiosensitivity of different tissues and the probability of damage should exposure to ionising radiation occur. Therefore effective dose is associated with stochastic effects:

$$Effective\ dose = \Sigma (equivalent\ dose \times tissue\ weighting\ factor) \ (Sv)$$

(T.1)

The tissue weighting factors are defined in the reports of the International Commission on Radiological Protection (ICRP). The most recent set of factors was published in ICRP Publication 103 and is shown in Table T.2.

The tissue weighting factor applied to the gonads has been reduced in recent years. This is due to research showing that the risk of hereditary disease from radiation of the gonads is not as high as was previously thought.

**Related Articles:** Absorbed dose, Radiation weighting factor, Equivalent dose, Effective dose, Stochastic effect, International Commission on Radiological Protection


### TLD (thermoluminescent dosimeter)

*(Radiation Protection)* See Thermoluminescent dosimeter (TLD)

### TOF (time-of-flight)

*(Magnetic Resonance)* See Time-of-flight (TOF)

### Toggle switch

*(General)* A toggle switch is an electrical switch that is operated by a mechanical lever. The lever usually has two separate stable positions and a snap action mechanism, so that a positive action occurs and it is easy to tell which position the switch is in. The switch is ‘toggled’ from one state to the other.

Such switches may act simply to make or break one pair of contacts (single pole, single throw), or switch between alternate contacts (single pole change over) – see Figure T.34. Multiple sets of contacts may be driven in concert to give a multi-pole switch.

### Tolerance

*(Radiotherapy)* Radiation treatment inevitably affects normal tissue and so may cause radiation induced adverse effects. The tolerance of normal tissues to radiation depends on the ability of the clonogenic cells to maintain a sufficient number of mature cells suitably structured to conserve organ function. The tissue architecture is thought to be important in determining the tolerance dose for partial organ irradiation. Groups of cells within an organ may

---

**TABLE T.2**

Tissue Weighting Factors as Defined in ICRP 103

<table>
<thead>
<tr>
<th>Tissue</th>
<th>Weighting Factor ($W_T$)</th>
<th>$\Sigma W_T$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bone marrow, breast, colon, lung, stomach</td>
<td>0.12</td>
<td>0.60</td>
</tr>
<tr>
<td>Bladder, oesophagus, gonads, liver, thyroid</td>
<td>0.05</td>
<td>0.25</td>
</tr>
<tr>
<td>Bone surface, brain, kidneys, salivary glands, skin</td>
<td>0.01</td>
<td>0.05</td>
</tr>
<tr>
<td>Remainder tissues</td>
<td>0.1</td>
<td>0.10</td>
</tr>
<tr>
<td><strong>Total</strong></td>
<td></td>
<td><strong>1.00</strong></td>
</tr>
</tbody>
</table>

---

**FIGURE T.33** Set up for measuring the tissue phantom ratio (TPR).

**FIGURE T.34** Various toggle switches.
be thought of as organised into collective bodies called functional subunits (FSU). If the integrity of a sufficient number of FSUs is maintained, the function of the organ is preserved. In some tissues, the FSUs are discrete, anatomically delineated structures with a clear relationship to tissue function. Examples of these structurally defined FSUs are the nephron in the kidney, the lobule in the liver, and the acinus in the lung. In other tissues, the FSUs have no clear anatomic demarcation. Examples of these structurally undefined FSUs include the skin, the mucosa and the spinal cord. These two types of tissue differ in their response to radiation.

For structurally defined FSUs their survival depends on that of one or more clonogenic cells within them and tissue survival depends on the number and radiosensitivity of these clonogens. Such tissues are composed of a large number of FSUs but each is considered a small self-contained entity independent of its neighbours. Surviving clonogens cannot migrate from one to the other. Consequently, survival of the FSU after irradiation depends on the survival of at least one clonogen within it. Taking the kidney and its FSU the nephron as an example, the survival of a nephron after irradiation depends on the initial number of renal tubule cells per nephron and their radiosensitivity. Since this FSU is relatively small, it is completely depleted of clonogens by low doses which accounts for the low tolerance to radiation of the kidney.

For structurally undefined FSUs the clonogenic cells are not confined to one particular FSU but can migrate from one to another allowing the repopulation of a depleted FSU following irradiation. For example, re-epithelialisation of a denuded area of skin can occur either from surviving clonogens within the denuded area or by migration from adjacent areas.

In radiotherapy, it is generally the case that the total dose that can be tolerated depends on the volume of tissue irradiated – the dose–volume effect. The spatial arrangement of the FSUs in the tissue is critical. In serial organs, the FSUs are arranged in series, like the links of a chain, and the integrity of each is critical to organ function. Damage to a single FSU is sufficient to cause a complication. Radiation damage to such tissues is expected to show a binary response: normal function for doses below a threshold dose above which there is loss of function. For these tissues, the greater the volume of tissue irradiated, the steeper the sigmoid dose–response curve becomes and the threshold dose decreases. This explains the volume effect observed in spinal cord, a serial organ where the loss of any one FSU may result in myelopathy.

In parallel organs, the FSUs are arranged in parallel so that the inactivation of a small number of FSUs does not lead to loss of organ function. Inactivation of a critical number of FSUs is required for functional damage to occur meaning that there should be a threshold volume of irradiation below which no functional damage will develop even after high-dose irradiation. Above this threshold there is a graded rather than a binary response: functional impairment increases in severity with increasing dose. Tissue with structurally undefined FSUs such as skin and mucosa respond in a similar way to tissues with a parallel FSU structure.

New knowledge of the dose–volume relationship for tissue tolerance is emerging from studies of patients who have received IMRT treatment. Traditional radiotherapy treatments generally give high doses to relatively small volumes of normal tissue whereas the highly conformal nature of IMRT results in large volumes of normal tissue receiving a low dose. There is increasing evidence that for some organs their tolerance to high doses may be affected by the volume receiving a low dose. Reviews of data for patients who received IMRT treatment for mesothelioma and for non-small-cell lung cancer indicate that the risk of fatal pulmonary events after radiotherapy may be due to relatively high lung doses (20 Gy) superimposed on extensive volumes of lung exposed to low radiation doses. For example, Yorke et al. report pulmonary complications at the 20% level occurred when >50% of the lung volume received ≥10 Gy. Such findings are not confined to parallel organs; experiments on rat spinal cord have shown that the radiation dose required to produce necrosis over short cord segments is significantly reduced if the adjacent tissue is first exposed to sub-threshold doses of radiation as low as 4 Gy. The suggested mechanism is that a negative effect is exerted by the low-dose bath on the regenerative capacity of neurons within the high-dose field.

Curative radiotherapy usually involves treating the normal tissues immediately surrounding the tumour to the limit of tolerance and a much larger volume will receive a lower dose. Unfortunately, a significant number of patients subsequently relapse or develop nodal disease or new tumours within the previously irradiated region. Decisions regarding re-treatment options require consideration of the dose received by normal tissues from the original treatment, the extent to which the normal tissues have regenerated, and on the extent of any residual or latent damage present after regeneration. In general, acutely responding tissues, such as skin and intestine, recover from radiation injury rapidly and can be re-irradiated to full tolerance within 1–3 months. However, for late toxicity endpoints tissues vary considerably in their recovery capability. The heart, bladder and kidney do not exhibit long-term recovery at all. In contrast, the skin, mucosa, lung and spinal cord are capable of limited long-term recovery and can be re-irradiated with partial tolerance doses. The extent of recovery is dependent on the organ type, magnitude of the initial radiation dose, and partly on the interval between radiation treatment courses.

**Abbreviations:** FSU = Functional sub-unit and IMRT = Intensity-modulated radiotherapy.

**Related Articles:** Adverse effect, Cell survival, Dose–response model, Intensity-modulated radiotherapy (IMRT), Normal tissue toxicity, Parallel organs, Serial organs, Sigmoid dose–response curve, Therapeutic ratio


**Tomography**

(General) Tomography is the process of imaging sections through a three-dimensional object. It is distinct from projection methods in which the effects of all tissue perpendicular to the imaging plane accumulate,
leading to the superposition of imaged tissues. Tomography derives from the Greek word *tomos* (section) and came into use in the field of x-ray imaging in the 1930s. The means of acquiring sectional images (tomograms) differs widely between imaging modalities, though there is substantial overlap between reconstruction methods.

**X-Ray:** Whereas projection images such as the skull x-ray in Figure T.35 show the accumulated absorption of x-rays between the source and the detector, tomograms show the absorption only of a slice of tissue of a specified thickness, allowing visualisation of structures in thin sections. At its inception, x-ray tomography was achieved by moving the x-ray source and film in opposite directions. Signal from tissue outside the focal plane was blurred, leaving a clear image of a section in the focal plane, the depth and thickness of which could be determined by the motion of source and film. In modern computed tomography (CT), sections through the patient are reconstructed from a number of projections acquired by rotation of the x-ray gantry around the patient axis, as in the axial CT of the head in Figure T.36. A number of reconstruction methods are available, including filtered back projection and iterative reconstruction. CT tomography was originally used to yield axial sections from oblique sagittal/coronal projections, as in the example in Figure T.36, but the near-isotropic resolution of modern CT scanners allows images to be extracted from reconstructed 3D image volumes in any orientation desired.

**Positron Emission Tomography:** Positron annihilation yields a pair of back-to-back 511 keV photons which are registered in a pair of detectors, indicating a line along which the event took place (a line of response). Annihilation photons are collimated so that only those travelling perpendicular to the patient axis reach detectors. Many such events build up projections of the distribution of radiotracer similar to the projections of x-ray absorption in CT. Similar reconstruction techniques are applicable, though the higher noise characteristics of PET make iterative reconstruction
with expectation-maximisation algorithms generally preferable to filtered back projection.

**Magnetic Resonance Imaging:** The term ‘tomography’ is seldom used in MRI except as a synonym for imaging because nearly all methods yield sections and projections are rarely reconstructed. In most MRI techniques two-dimensional or three-dimensional images are computed using Fourier transformation of data that is sampled in a Cartesian fashion from selectively excited 2D sections or 3D volumes. In a small number of techniques such as projection-reconstruction imaging, the space of the measured signal (k space) is sampled radially rather than in a grid so as to generate projections. These can be reconstructed into tomograms with similar approaches to those used in CT and PET.

**Ultrasound Imaging:** Ultrasound imaging is an inherently tomographic method in which the profile of the ultrasound beam from the probe defines the thickness of the imaged section.

**Abbreviations:** CT = Computed tomography, MRI = Magnetic resonance imaging and PET = Positron emission tomography.

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**Tomosynthesis**

*(General)* Tomosynthesis is the process of using images of multiple projections of an object to construct images of planes through that object. In conventional x-ray focal plane tomography a single exposure is taken while a tube and film move about a fulcrum in a chosen object plane. This produces a clear image only of that plane – all other planes are blurred in the film to some degree. The objective of tomosynthesis is to form an image of any plane in the object by processing images of multiple projections of the object. Grant (1972) was the first to adopt the term tomosynthesis and presented an analog method of taking and combining radiographs to view any plane through an object. While focal plane x-ray tomography was a routine, feasible examination with analog techniques and detectors, multi-projection tomosynthesis was not.

The introduction of flat panel detectors and digital image processing techniques had led to a renewed interest in tomosynthesis. The low geometric distortion and high sensitivity of flat panel detectors have made tomosynthesis at relative low doses practicable. In mammography in particular the technique appears promising (Diekmann and Bick, 2007). In conventional mammography, evidence of pathology can be obscured, particularly in dense breast. Several reconstruction algorithms have been used, including matrix inversion, filtered backprojection and maximum likelihood approaches (Rakowski and Dennis, 2006).

The technique is also finding applications in musculoskeletal imaging and in chest imaging, clarifying pulmonary nodules that may be obscured in a conventional chest x-ray (Vikgren et al., 2008), without the need to resort to a higher dose CT exam.


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**Tomotherapy**

*(Radiotherapy)* Tomotherapy is an external beam radiotherapy technique that delivers the dose distribution using similar geometry to a tomographic CT scanner (hence the name tomo). Two variations on the geometry have been developed. The first is helical tomotherapy, which mimics a helical CT scanner’s geometry and is a continuous irradiation with the linear accelerator revolving around the patient, delivering a fan beam of radiation, as the patient position is slowly indexed. The second is slice-by-slice tomotherapy, in which a CT-image-like slice is irradiated with a fan of radiation from the linear accelerator, and then the patient position is indexed to treat the adjacent region. This is repeated until the whole target volume is irradiated. For both implementations a one-dimensional binary collimator is used to shape the dose distribution as the beam rotates. This binary collimator has a set of high attenuation collimating fingers which are either fully in or fully out of the beam. In the case of slice-by-slice tomotherapy, two banks of binary collimators are used to allow two slices to be treated simultaneously. There are two main tomotherapy manufacturers. Tomotherapy Inc. makes a helical tomotherapy system with the product name Tomotherapy. Nomos makes a slice-by-slice system with two banks of collimators called the MiMİC. Tomotherapy is often used as a specific product name for the helical system and as a general term for this class of technology. Often tomotherapy is accompanied by megavoltage conebeam CT (MVCT) using the treatment beam to enable imaging during treatment.

Alternatives to tomotherapy for external beam treatment are the most common technique using a convention set of fixed beam angles which are shaped in two dimensions, arc therapy with a conventional treatment unit and collimation system, and intensity-modulated arc therapy (IMAT) which combines conventional arc therapy with beam shape and dose rate modulation.

**Abbreviations:** CT = Computed tomography, IMAT = Intensity-modulated arc therapy and MVCT = Megavoltage CT.

**Related Articles:** Linear accelerator, Computed tomography, Intensity-modulated radiotherapy, Image-guided radiotherapy, Intensity-modulated arc therapy, Arc therapy


**Hyperlinks:** The Nomos corporation: [http://www.nomos.com](http://www.nomos.com/); Tomotherapy Inc: [http://www.tomotherapy.com](http://www.tomotherapy.com/)

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![FIGURE T.37](image.png) Breast tomosynthesis. Multiple projections of the breast are taken as the tube moves in an arc.
Tongue and groove leakage

(Radiotherapy) Adjacent MLC leaves must slide smoothly across each other with minimal gaps between leaves to reduce leakage and transmission radiation. This is accomplished by machining tongue-and-groove patterns into the sides of the leaves. The side of one leaf has an extended portion called the tongue, while the abutting side of the adjacent leaf has an indented portion called the groove. Two adjacent leaves are coupled together as the tongue of one leaf slides within the groove of the adjacent leaf. Each leaf has a tongue on one side and a groove on the opposite side. The manufacture of precise tongues and grooves into extremely small individual leaves is an exacting process that places a lower limit on the size of individual MLC leaves (Figure T.38).

While this tongue-and-groove design reduces radiation leakage, it also complicates treatment planning dose calculation because the transmission through any leaf depends on whether the beam passes through the tongue, the centre or the groove portion of the leaf. There are some interleaf effects falling into two categories: interleaf leakage and tongue-and-groove effect. When adjacent leaves travel in synchrony across the field, the transmission through the narrow gap between the adjacent leaves is larger than the transmission through the middle of the leaf (interleaf leakage). But when the adjacent leaves are not synchronized, irradiation occurs sequentially through the tongue of one leaf and the groove of the adjacent leaf, whereby the dose in the region between the adjacent pairs of leaves is less than that of the average of the two pairs of leaves (tongue-and-groove effect).

The amount of interleaf leakage can be easily measured using some radiographic film, and should typically be less than 3% of the dose in the beam. A film is placed at the isocentre on the couch and one field is delivered on each half of the film. Firstly a rectangular open field is delivered to one side of the film with a sufficient number of monitor units set to yield a useable optical density. Then the couch is moved laterally to avoid overlap of the two fields prior to the second field. This second field is the same size but has one bank of MLC leaves covering the open portion of the field, with no additional shielding behind the MLCs. An illustration of the fields is shown in Figure T.39. A much larger number of monitor units is set (approximately 25 times the first field) for this field to allow a sufficiently significant optical density to be recorded on the film. The maximum optical density for each field is recorded (for the MLC field the transmission is measured along a line approximately through the middle of the leaves – see Figure T.39) and when the optical density of this field is read out the value is then divided by 25. The procedure is repeated for the opposite bank of MLC leaves.

Another means of reducing this leakage is to ensure that the back-up jaws (secondary collimators) are positioned as close as possible to, but not within, the field edge. This will provide a further approximate ten-fold reduction in dose; therefore the shielded areas should receive a dose of around 0.5% of the dose to the open field area.

Abbreviation: MLC = Multileaf collimator.
Related Articles: Leakage radiation, Multileaf collimator

Topogram

(Diagnostic Radiology) Topogram is a vendor name (SIEMENS) for the scan projection radiograph used in computed tomography.
Related Article: Scan projection radiograph

Total body irradiation

(Radiotherapy) Total body irradiation (TBI), as its name implies, is a radiotherapy technique that involves irradiation of the whole body. It is often used prior to bone marrow or blood stem cell transplantation and its purpose is to suppress the recipient’s immune system preventing rejection of the transplanted bone marrow or blood stem cells. Doses used may be of the order of 12 Gy. Treatments are often fractionated but may be single fraction.

Total body potassium

(Radiotherapy) The human body comprises distinct and measurable body compartments. The status and rate of change of these compartments reflect the health of a person and the response of treatment for a specific disease. Whereas measurement of body weight is a basic and useful parameter, it can be a misleading measure of disease status and response to treatment, which may increase oedema and fat while depleting protein.

Total body potassium (TBK) is one such compartment that provides information on the fat free mass (FFM), as potassium is not found in adipose tissue and the fat-free tissues of the body contain a constant proportion of potassium.

TBK is associated with the metabolising, oxygen-consuming portion of the body, so a decline in TBK is usually interpreted as a loss of muscle mass due to a catabolic condition. However, TBK is not a gold standard for the measurement of fat free mass in sick subjects, as changes in TBK can reflect chemical changes in the body and blood.

Whole Body Counting is the measurement of induced or naturally occurring radiation in the body. It is performed in a shielded room or chamber with low background radiation. The background radiation is subtracted from the total radiation measured. A whole body counter (WBC) is used to measure the total body potassium in human subjects, and radioactive contamination in accident conditions.

Potassium–40 has a half life of 1.3 x 10^9 years, and emits a 1.46 MeV gamma ray on decay. ^{40}\text{K} has an abundance of 0.0117% in natural potassium. The average potassium content of an adult male is ~140g, or about 15mg of ^{40}\text{K}. This translates to an emission rate of 30,000 photons/min. In adult females, the level is about 20,000 photons/min.

A 1.46 MeV decay gamma ray is readily detected in supine geometry by an array of NaI detectors. This measurement does not give the body any additional radiation dose, as the activity is already present in all cells in the FFM. A low background, shielded room is constructed with old steel, forged prior to the A-bomb testing program. The precision and accuracy of this method, expressed as coefficients of variation, are 1.5% and 4.5% respectively. FFM is calculated from TBK on the basis that the potassium content of FFM is 2.26g/kg in females and 2.52g/kg in males.

Abbreviations: FFM = Fat free mass, TBK = Total body potassium and WBC = Whole body counter.

Related Articles: In vivo body composition, Total body protein, Total body nitrogen, Total body fat

Total body potassium

(Radiotherapy) The human body comprises distinct and measurable body compartments. The status and rate of change of these compartments reflects the health of a person and the response of treatment for a specific disease. Whereas measurement of body weight (M) is a basic and useful parameter, it can be a misleading measure of response to treatment, which may increase oedema and fat while depleting protein.

An important compartment is the total body protein (TBP), which is a measure of muscle and visceral mass. It is determined directly by measurement of the total body nitrogen (TBN):

\[
\text{TBP} = 6.25 \times \text{TBN}
\]

TBN is regarded as the superior measure of protein status in diseased subjects.

Methods: In vivo methods are used to obtain values of the body compartments. These might be done before and after therapy, to determine the effect of therapy. The measurement of normal values is required so as to compare patient values, according to defined index. The nitrogen index \( \text{NI} = p\text{TBN}/n\text{TBN} \) where \( p \) designates the patient and \( n \) the normal values for healthy subjects.

The importance of such indices relates to the impact on disease prognosis.

In vivo interrogation techniques are used. These can give a measure of the whole compartment, or in some cases the spatial distribution of the compartment.

Nitrogen is measured in a body protein monitor (BPM) using a Cl252 or PuBe neutron source. The patient is moved over a collimated neutron beam. The fast neutrons emitted by these radioactive sources are moderated in tissue and captured by nitrogen and hydrogen nuclei in the patient. The 11.4 MeV ground state gamma ray can be measured by NaI detectors, being of higher energy than all background radiations. The hydrogen gamma ray is very intense and easily detected. The ratio of nitrogen to hydrogen yields eliminates the dependence of gamma ray attenuation on body habits.

Abbreviations: BPM = Body protein monitor, \( \text{NI} = \) Nitrogen index, \( \text{TBN} = \) Total body nitrogen and \( \text{TBP} = \) Total body protein.

Related Articles: In vivo body composition, Total body nitrogen, Total body potassium, Total body water, Total body fat

Total body water

(Radiotherapy) The body comprises distinct and measurable body compartments. The status and rate of change of these compartments reflects the health of a person and the response of treatment for a specific disease. Whereas measurement of body weight (M) is a basic and useful parameter, it can be a misleading measure of response to treatment, which may increase oedema and fat while depleting protein.

The mass of a subject is the sum of the defined compartments in the four body compartment model:

\[
\text{M} = \text{TBP} + \text{TBK} + \text{TBW} + \text{TBCa}
\]

Methods: Analysis of blood and urine after ingestion of isotopically labelled molecules is used as the gold standard. Heavy water \((\text{D}_2\text{O})\) is ingested and after 12h nil by mouth, blood or urine samples are taken and analysed for deuterium by Fourier transform infrared analysis. This is a low-cost and accurate dilution technique to determine total body water.

Abbreviations: TBCa = Total body calcium, TBF = Total body fat, TBP = Total body protein and TBW = Total body water.

Related Articles: In vivo body composition, Total body protein, Total body nitrogen, Total body potassium, Total body fat

Total brightness gain

(Diagnostic Radiology) There are a number of parameters used to assess an image intensifier (II).

Total brightness gain is one of the main parameters of II. It is defined as

\[
\text{Total brightness gain} = (\text{output light photons})/(\text{input x-ray photons})
\]

Usually this figure is between 1000 and 6000.

Another parameter of II is the minification gain (the ratio between the input and output screens – area or diameter):

Minification gain \((\text{Gm}) = (\text{area of input screen})/(\text{area of output screen}) = (D_{\text{inp}}/D_{\text{out}})^2\)


This figure will depend on the diameters, for example 12 in. image intensifier with 1 in. output screen diameter will have \((12/1)^2 = 144\) millification gain.

Another parameter of an image intensifier is the flux gain, also known as electronic gain (the ratio between the internal light photons in the image intensifier):

\[
\text{Flux gain (GF)} = \frac{\text{Output screen light photons}}{\text{input light photons to the photocathode}}
\]

Usually this figure is between 30 and 60, most often around 50. The product of Gm and Gf is known as total gain \((G)\):

\[
G = \text{Gm} \times \text{Gf} \quad \text{(from the figures above } G = 144 \times 50 = 7200)\]

A parameter which links the radiation dose used to produce the image and the output light is the conversion factor:

\[
\text{Conversion factor} = \frac{\text{output phosphor light}}{\text{input screen dose rate}}
\]

Usually this figure is between 100–1000 \((\text{cd/m}^2/\mu\text{Gy/s})\).


Total scatter factor
(Radiotherapy) The total scatter factor is the product of the collimator scatter factor (CF), and the scatter factor (SF). See related articles for more information.

It is also known as the relative dose factor (RDF). It can be calculated directly as the ratio of the dose in a phantom, field size \(A\), to the dose in a phantom for a \(10 \times 10\text{ cm}^2\) field.

Related Articles: Scatter factor, Collimator scatter factor

Total skin irradiation
(Radiotherapy) Total skin irradiation (TSI) is an electron-based treatment technique. It has been used since the 1950s and is used to treat diseases including: mycosis fungoides, which is a common chronic form of cutaneous T-cell lymphoma; leukemia cutis; Kaposi’s sarcoma and scleromyxoedema. This is a large SSD irradiation technique and often used electron energies of 3–10 MeV.

Abbreviations: SSD = Source to skin distance and TSI = Total skin irradiation.

Related Article: Electron therapy

TR (repetition time)
(Magnetic Resonance) See Repetition time (TR)

Trace of the diffusion tensor
(Magnetic Resonance) The trace of the diffusion tensor (described by a \(3 \times 3\) square matrix) is the sum of its diagonal elements. Equivalently, the trace can be defined as the sum of the eigenvalues of the diffusion tensor. The trace is thus invariant with respect to a change of basis, that is the directions in which the apparent diffusion coefficients (ADC) are measured.

The trace of the diffusion tensor can be estimated as the sum of the ADCs in three orthogonal directions. Moreover, the mean diffusivity (MD) is a third of the trace of the diffusion, since MD is given by the average of the eigenvectors of the diffusion tensor.

Related Articles: Apparent diffusion coefficients (ADC), Eigenvalues, Mean diffusivity, Diffusion tensor

Tracer
(Nuclear Medicine) Tracer is a common name to describe a substance that when injected into the body follows ‘traces’ a certain or a chain of physiological or biochemical processes. In nuclear medicine radioisotopes are often labelled to tracers. These compounds are referred to as radiotracers or radiopharmaceuticals. The temporal distribution of the radionuclide can be measured in a dynamic study.

There are three different types of tracers: (1) naturally occurring substances that are suitable for labelling; (2) substances that are analogs of natural substances, that is mimic the behaviour of naturally occurring substances and (3) compounds that participate and interact with different physiological or biochemical processes in the body.

An ideal tracer should fulfill the following properties:

1. Tracer behaviour should be identical or similar to the natural substance.
2. The mass of the tracer should be small relative to the endogenous compound being traced. Typically the tracer mass should be <1% of the endogenous compound.
3. A high specific activity is required to allow imaging and blood or plasma activity assays.
4. The consequence of the isotope effect (see Related Article) should be negligible or quantitatively predictable.

In some cases when the tracer is labelled with an element that is not originally present in the compound, the tracer must still behave in a way that is similar to the natural substance. \(^{99m}\text{Tc}\), \(^{121}\text{I}\) and \(^{18}\text{F}\) are examples of radioisotopes which are not normally present in biological systems but are commonly used when labelling compounds. Such radionuclides can be used to monitor simple parameters that are related to distribution, transport and excretion, but since they are not normally present in human biochemistry (iodine excepted, when studying thyroid metabolism), they are unsuitable for tracing biochemical reactions in the body. These biochemical systems are more specific than the transport system. When a foreign element is introduced into a compound, its biochemical properties are likely to change. Such a radionuclide is not a good representation of the biochemical processes it is required to study. Radionuclides which represent elements that normally take part in biological processes, such as \(^{14}\text{C}\), \(^{13}\text{N}\) and \(^{15}\text{O}\), have the advantage that they generally do not alter the behaviour of the labelled compound. On the other hand these radionuclides might have undesired physical characteristics or an expensive production cost, which limits their usefulness.

When using analog tracers it is possible to tailor the compound so that the tracer only participates in chosen parts of the biochemical sequence. This is a desired feature when trying to decrease the number of variables modelled in the activity time curve. One example of an analogue tracer is FDG (\(^{18}\text{F}\)-2-fluoro-2-deoxy-d-glucose) which measures glucose metabolism.

Related Articles: Isotope effect, Analogue tracers


Tracer delivery
(Nuclear Medicine) Tracer delivery refers to the process in which a tracer is extracted from a tracer transport system, for example the delivery of oxygen via the blood passing through the capillaries. A number of factors determine the magnitude of the extracted
fraction, namely the blood flow $F$ and the extraction and clearance of tracer from the blood.

Blood flow can be measured as volume per unit time or as volume per unit time per mass unit. The latter is referred to as perfusion or blood flow per unit tissue mass.

The extraction is defined as net and unidirectional extraction. The net extraction depends on the difference in steady state tracer concentration in the blood entering the compartment (arterial) $C_A$ and the blood leaving the compartment (venous) $C_V$. The net extraction $E_n$ is defined by

$$E_n = \frac{(C_A - C_V)}{C_A} \quad (T.2)$$

If there is no metabolism of the tracer, all of the tracer will be returned to the blood and the net extraction will equal zero. This could be the case when using inert diffusable blood flow tracers when a steady state is reached.

The unidirectional extraction on the other hand refers only to the tracer uptake to tissue from the blood and does not account for the back-transfer of the tracer.

The Fick principle is used to correlate the processes of blood flow, flux and extraction, and it states that, under steady state conditions, the extraction of tracer from the blood is equal to the difference in tracer concentration in blood input (arterial blood before the organ capillaries) and output (venous blood after the organ capillaries). The uptake rate $U$ (mg/min) is determined by the blood flow $F$ (mL/min), the arterial tracer concentration $C_A$ and the venous tracer concentration $C_V$. The uptake rate is

$$U = F \times (C_A - C_V) \quad (T.3)$$

The Fick principle can only be applied on a steady state system.

The amount of tracer leaving the blood depends on the extraction through the capillary walls or membrane. The amount of tracer extracted depends on the capillary surface area $S$, the capillary permeability for the tracer $P$, and the blood flow $F$. The relationship between these quantities can be described by a simple model proposed by Renkin and Crone. The model is based on four assumptions; (1) an idealised capillary (i.e. rigid tube), (2) tracer concentration is constant throughout the capillary, (3) the extraction of tracer from the blood depends on the blood tracer concentration and (4) there is no back-transfer of the tracer, that is unidirectional blood flow. The extraction fraction for such a system is determined by

$$E_n = 1 - e^{-\left(\frac{P \times S}{F}\right)} \quad (T.4)$$

The Renkin-Crone model is not completely realistic but it provides instructive illustrations about the relationship between the extraction fraction and the blood flow, permeability of the capillary and the capillary surface area. The ratio ($P \times S/F$) is called the extraction coefficient. The permeability-surface product ($P \times S$) can be considered as a representation of the tracer flow from the blood to the tissue through the capillaries, while $F$ is considered the actual blood flow. Given by ($P \times S/F$) is that the extraction fraction can be increased by either reducing the blood flow through the capillaries ($F$) or by increasing the flow through the capillary walls ($P \times S$).

But at the same time the amount of tracer extracted to the tissue also depends on the product of the extraction fraction $E$ and the blood flow $F$. This product is also referred to as clearance. For example, during exercise the heart beats faster, that is increasing the blood flow, to provide the muscles with more oxygen and nutrients. The extraction fraction decreases due to the increase in blood flow but at the same time more nutrients and oxygen reach the capillaries which allows for a greater transport. This increase in capillary transport often more than offsets the decrease in the extraction fraction. The tracer clearance from blood to tissue is used in tracer kinetic modelling.

**Related Articles:** Tracers, Analogue tracers, Distribution volume, Partition coefficient


**Tracer flux between compartments**

*(Nuclear Medicine)* The amount of substance that crosses a boundary per unit time is referred to as flux. The boundary can be a membrane or a cross section of a blood vessel. Flux also refers to the transport of tracer between two compartments. The unit for tracer flux is flux per unit volume or mass of tissue (e.g. mol/min/mL or mg/min/g).

**Rate Constant:** The substance flux between two compartments is described by a number of rate constants $k_i$. In a simple first-order process the flux is the product of the rate constant and the amount (e.g. concentration or mass) of a substance in a compartment:

$$\text{Flux} = k \times \text{Amount of substance in compartment} \quad (T.5)$$

The unit of $k$ is $(\text{time})^{-1}$. Depending on whether ‘amount’ refers to mass or concentration the unit is mass/time (mg/min) or mass/time per unit of compartment volume (mg/min/mL) respectively. If the rate constant is 0.1 min$^{-1}$ it would mean that the 10% of the substance is transported out or in per minute. The inverse of the rate constant is referred to as the turnover time or mean transit time $\tau (k = 0.1 \text{ min}^{-1} \Rightarrow \tau = 10 \text{ min})$. The turnover is described by an exponential function analogous to the radioactive decay factor, and the time it takes for the original amount of tracer to decrease by 50% (assuming no back transfer), the halftime of turnover, is given by

$$t_{1/2} = \frac{0.693}{k} \quad (T.6)$$

In most situations there is more than one potential pathway for a tracer out from the compartment, each associated with an individual rate constant $k_i$. The half time of the turnover in such a situation is the inverse of the sum of the rate constant

$$t_{1/2} = \frac{0.693}{(k_1 + k_2 + \cdots + k_m)} \quad (T.7)$$

where $m$ is the number of possible tracer pathways out of the compartment.

The most common compartment models are based upon the assumption that the dynamics of the system are based upon first-order kinetics, that is linear behaviour of the tracer kinetics. For example, when doubling the concentration the flux is also doubled. These models also adequately model compartments with non-linear tracer kinetics.

**Related Articles:** Tracers, Analogue tracers, Distribution volume, Partition coefficient, Steady state condition in tracer kinetic modeling

Tracer kinetic modelling

(Nuclear Medicine) Tracer kinetic modelling uses mathematical models that describe the temporal behaviour and distribution of radiopharmaceuticals. The modelled behaviour of a tracer is used to examine biological processes and determine radiation doses to specific organs. The temporal and spatial distribution of an injected radiotracer in any specific organ depends on the tracer kinetic characteristics; tracer delivery, binding to cell surface receptors, diffusion and transport through cell membranes, metabolism, clearance from cells, wash out from the tissue and excretion from the body.

Tracer concentration in an organ is modelled over time by a *time activity curve*. Dynamic studies can provide information about the biological fate of the tracer after injection. The conclusions drawn from these studies can be used in mathematical models with a number of parameters explaining the time activity curves. Some of these parameters can be directly related to biological or physiological processes, for example tissue perfusion.

**Related Article:** Time activity curve


Tracer transport

(Nuclear Medicine) Tracer transport refers to the transport of tracer across barriers in a biological system.

There are three different ways of transporting a tracer across a membrane or capillary wall, namely active transport and two forms of passive transport: passive diffusion and carrier-mediated diffusion.

*Active transport* involves a process that requires energy. Substances in an active transport can move against concentration gradients. The energy source for an active transport is often adenosine triphosphate (ATP). The sodium-potassium pump and the renal tubular reabsorption of glucose are two examples of active transport.

*Passive transport* on the other hand does not require energy and will only move in the same direction as the concentration gradient. *Passive diffusion* through a membrane or a capillary is described by a *diffusion constant* $D$ (cm$^2$/min). The permeability $P$ (cm/min) of the membrane is related to the diffusion constant according to

$$p = \frac{D}{x}$$

where $x$ is thickness of the membrane. A large $D$ represents a rapid clearance of tracer from the blood via passive diffusion mechanisms. A large number of processes and substances depend on passive diffusion, for example water, oxygen, ammonia and carbon dioxide.

Transport using carrier-mediated diffusion involves a carrier molecule $C$ and a substrate molecule $S$, for example when transporting glucose and amino acid over the blood brain barrier. The substrate, that is the tracer, is transported to the membrane via the vascular system, and at the membrane it forms a carrier-substrate complex $SC$ that moves across the barrier before it disassociates into its original compounds $S$ and $C$ again.

The substrate transport rate is proportional to the amount of substrate, but since there are only a limited number of carrier molecules available the transport can be saturated. A common carrier is a protein enzyme which is neither created nor destroyed in the process but works as a catalyst, that is speeding up the process.

**Related Articles:** Analogue tracer, Distribution volume, Partition coefficient, Tracer flux between compartments, Tracer kinetic modelling


Tracking system

(General) A tracking system is a device that determines and tracks the position of a patient, organ or device within the patient environment.

This may be used for a variety of purposes, and forms the basis of image-guided therapies and for automatic detection and correction of patient movement in diagnostic imaging and therapy.

A wide range of tracking techniques are available, some relying only on the patient’s external features, others on passive external markers, whilst more sophisticated devices may rely on active external markers or even passive or active internal markers.

The common process is to detect known points or markers and determine their positions relative to a datum in the room, and repeat the process often enough to accurately ‘track’ any movement.

**Marker tracking systems include**

- Passive – IR sensitive multi-camera detection of reflective markers
- Active – Fixed multi-sensor detection of IR light emitting diode markers
  - Fixed magnetic sensors detecting signals from powered marker coils
  - Fixed microphones detecting sounds from spark gap markers

Surface tracking systems rely on computer image processing of one or more camera images of the patient by

- Projecting a complex pattern of light onto the patient’s surface and analysing the observed images and calculating the 3D surface that best fits the image observed by one or more cameras
- Comparing a 2D camera view image against a previously determined 3D dataset of the patient’s shape (e.g. from reconstructed MRI data)

Inherent tracker systems are also being developed which use the 3D medical imaging or therapy device itself to image the patient and track the organ of interest or markers. In these cases, sophisticated fast image processing and segmentation algorithms are necessary to identify the organ boundary or marker positions and track movement.

Tractography

(Magnetic Resonance) Tractography is a method used to construct and visualise neuronal tracts or other fibre bundle. It is based on the main direction of the diffusion tensor derived from DWI studies of the water self-diffusion in three dimensions. The method relies on the assumption that water diffusion is faster along than perpendicular to the fibre bundles. By statistical analysis of the principal diffusional direction in adjacent image voxels 2D trajectories in 3D space are constructed. The analysis is a seed-based tracking, where the fractional anisotropy value is used above a threshold for determining for how long the paths should be followed, and the angle between adjacent principal eigenvectors is used to determine whether the path is proper or not.

The technique can be used for characterising nerve fibre disturbances diseases like tumours or neurodegenerative diseases such as multiple sclerosis (MS), Parkinsonian disorders, or various types of dementia.

In Figure T.40 a tractography projection is shown, indicating the principle diffusion direction which is taken to correspond to the orientation of nerve fibres.
Training

Training is the activity leading to acquiring certain competencies (skills) necessary to practice a profession. While education equips the learner with the necessary academic knowledge (most often through a university course), training equips the learner with the vocational knowledge (most often through a practical course).

Medical physics training follows specific rules and requirements, which can build certain listed competencies. This training can be combined with education (through a parallel university course), but is better to follow the education. Various countries have different requirements for training. In some countries a post-education training course is the standard way for a person to reach state registration/certification as suitable to perform independently his/her professional tasks.

Large professional bodies have their own training requirements and schemes. For example, the UK Institute of Physics and Engineering in Medicine (IPEM) has a training scheme which is maintained (and regularly updated) by one of its committees. The training materials EMERALD and EMIT list specific training tasks, which will help the trainee to acquire specific competencies and vocational skills. These can be seen in the training tables of these materials (available in each demo of the materials).

Hyperlinks: [www.ipem.ac.uk](http://www.ipem.ac.uk); [www.emerald2.eu](http://www.emerald2.eu).

Transducer

(Ultrasound) An ultrasonic transducer acts as a transmitter, receiver or, in pulse–echo systems, as both. It converts electrical energy to acoustical and acoustical energy to electrical.

A simple single element transducer is shown in Figure T.41 with a circular piezoelectric disc, connected with electrodes on each flat end. A damping material can be mounted on the rear of the disc and a matching layer on the front side. The disc will vibrate (produce sound) if an electrical pulse is applied between the wires and conversely an electrical signal will occur if a sound wave hits the disc and makes it vibrate, Figure T.42.

PZT, lead zirconate titanate, is the most commonly used transducer material. This material is a synthetic ceramic which can be produced in various versions for different applications requiring particular properties. It consists of small crystals with separated positively and negatively loaded parts. When the material is compressed, this neutral load distribution is disturbed and a voltage across the flat ends occurs.

The transducer disc resonates best when applying a signal with a frequency corresponding to a wavelength of \( n \times \frac{d}{2} \), with \( d \) as the thickness of the disc, which means that it is more sensitive and has

**FIGURE T.40** Tractography.

**FIGURE T.41** Principle design of a single element ultrasound transducer. (Graphs courtesy of EMIT project, [www.emerald2.eu](http://www.emerald2.eu))

**Training**

(Ultrasound) An ultrasonic transducer acts as a transmitter, receiver or, in pulse–echo systems, as both. It converts electrical energy to acoustical and acoustical energy to electrical.

A simple single element transducer is shown in Figure T.41 with a circular piezoelectric disc, connected with electrodes on each flat end. A damping material can be mounted on the rear of the disc and a matching layer on the front side. The disc will vibrate (produce sound) if an electrical pulse is applied between the wires and conversely an electrical signal will occur if a sound wave hits the disc and makes it vibrate, Figure T.42.

PZT, lead zirconate titanate, is the most commonly used transducer material. This material is a synthetic ceramic which can be produced in various versions for different applications requiring particular properties. It consists of small crystals with separated positively and negatively loaded parts. When the material is compressed, this neutral load distribution is disturbed and a voltage across the flat ends occurs.

The transducer disc resonates best when applying a signal with a frequency corresponding to a wavelength of \( n \times \frac{d}{2} \), with \( d \) as the thickness of the disc, which means that it is more sensitive and has

**FIGURE T.42** Piezoelectric effect. (Graphs courtesy of EMIT project, [www.emerald2.eu](http://www.emerald2.eu))
The transformer is step-up if the windings of the secondary coil are more than those of the primary (e.g. 50,000 turns in the secondary coil and 100 windings in the primary coil, producing windings ratio of 500). This ratio is typical for a high voltage x-ray transformer and will produce 110kV from a primary voltage of 220V (220 × 500 = 110,000).

The transformer is step-down if the windings of the secondary coil are less than those of the primary (e.g. 10 turns in the secondary coil and 100 windings in the primary coil, producing windings ratio of 1/10). This ratio is typical for the filament transformer of an x-ray transformer and will produce 22V from a primary voltage of 220V (220 × 0.1 = 22).

See a diagram with both types of transformers in the article High voltage generator.

There are various types of transformers according to the shape of the ferromagnetic core. Some more common types are as follows:

- Air core transformer – This transformer has no iron core. It consists of simply two insulated primary and secondary coils placed closely one to the other.
- Open core transformer – In this transformer the primary and secondary coils are each separately wrapped around their own ferromagnetic cores. Again both coils + cores are placed closely to each other. Some autotransformers are open-core type transformers (both insulated coils are wrapped around a single iron core).
- Closed core transformer (most often used) – In this transformer the primary and secondary coils are wrapped around an iron core with closed geometry (either a ring or square shape).
- Shell-type transformer – The iron core of this transformer has three columns (in an E shape), which form two rectangular holes. The primary and secondary coils are wrapped around the central column. This way the core surrounds the coils. This design is used in some power transformers.

Transformers also vary according to the type of ferromagnetic material of the core – most often various types of iron core or ferrite core transformers.
Transformers also vary according to the phase of the AC current—most often single-phase transformers or three-phase transformers.

**Related Articles:** Air core transformer, Open core transformer, Transformer core

### Transformer core

*(Diagnostic Radiology)* The core of a transformer is made of ferromagnetic material with specific magnetic permeability \( \mu \) (e.g. \( \mu \) for ferrite materials is thousand times higher than \( \mu \) for iron).

The core can be of different shapes or sizes. For transformers with identical materials of the core, the larger is the required electrical power, the larger is the core of the transformer.

There is a known transformer relation between the transformer core and the frequency of the AC electricity used:

\[
\frac{U}{f} \sim An
\]

where
- \( A \) is the cross section of the transformer core (mm²)
- \( n \) is the transformer ratio (based on the number of secondary/primary windings)
- \( f \) is the frequency of the secondary voltage \( U \).

This relation is in the base of the contemporary high frequency x-ray generators. These use transformers with ferrite core, allowing high frequencies to be used. Due to this reason the size of the transformer core is approximately 25% of the size of similar high voltage step-up transformer with iron core (used in the classical x-ray generators). For more information see the article on *High frequency x-ray generators*.

### Transformer oil

*(Diagnostic Radiology)* Special insulation oil in which high voltage transformers are usually immersed. The specific insulation of the oil is normally more than 220kV/cm. The same oil is used inside the x-ray tube housing—both for insulation and cooling of the x-ray tube.

### Transformer winding

*(Diagnostic Radiology)* See article on Transformer

### Transient charged particle equilibrium

*(Radiation Protection)* When a photon beam impinges on a medium, the transfer of energy from the beam to the electrons does not lead to the absorption of energy exactly at the same location where the photon interaction takes place. This is due to the finite range of the secondary electrons released. In an ideal situation inside a medium, losses of energy due to electrons which leave the volume of interest are compensated by energy associated with incoming charged particles from surrounding parts of the medium. This is known as a charged particle equilibrium condition.

In a more realistic situation, that equilibrium is not generally achieved. At or near the entrance surface of the medium the absorbed energy (absorbed dose) is smaller than the transferred energy (kerma) because of the lack of equilibrium between the incoming and the exiting energy in the volume of interest. At a certain depth both magnitudes become equal. After that point where the maximum absorption is achieved, a transient charged particle equilibrium condition occurs: attenuation of the beam reduces its intensity and absorbed energy (absorbed dose) becomes greater than energy transfer (kerma). Figure T.45 illustrates those phenomena.

**Related Articles:** Charged particle equilibrium, Secondary electrons, Absorbed dose, Kerma

### Transient equilibrium

*(Nuclear Medicine)* Transient equilibrium refers to the activity equilibrium that occurs in a decay series when the half-life of the parent nucleus is approximately 10 times longer than the daughter nucleus half-life.

In the case of a transient equilibrium the relationship between the number of parent and daughter atoms \( (N_1 \text{ and } N_2 \text{ respectively}) \) is described by the following equation:

\[
\frac{N_2}{N_1} = \frac{\lambda_2}{\lambda_2 - \lambda_1}
\]

where \( \lambda_1 \) and \( \lambda_2 \) are the decay constants of the parent and daughter nucleus respectively.

In a technetium generator \(^{99}\text{Mo}\) decays to \(^{99m}\text{Tc}\) which is extracted as technetium-pertechnetate. \(^{99m}\text{Tc}\) is extracted when the \(^{99m}\text{Tc}\) activity reaches its maximum at time point \( t_{\text{max}} \):

\[
t_{\text{max}} = \left[ \frac{1.44 \cdot T_p \cdot T_d}{(T_p - T_d)} \right] \ln \left( \frac{T_p}{T_d} \right)
\]

where \( T_p \) and \( T_d \) are the half-lives of the parent and daughter nucleus respectively. \( t_{\text{max}} \) for \(^{99m}\text{Tc}\) is 22.9 h. (Figure T.46).

**Related Articles:** Secular equilibrium, Bateman equation for secular equilibrium, Bateman equation for transient equilibrium

### Transit time

*(Magnetic Resonance)* See Mean transit time (MTT)

### Transmission coefficient

*(Ultrasound)* When a propagating ultrasound wave encounters a medium with different acoustic impedance \( (Z) \) part of it will be reflected and the other part will be transmitted. The definition of the pressure transmission coefficient is (Figure T.47).

\[
T_p = p/p_i\text{, and can be expressed as } T_p = 2Z_i(Z_r + Z_i)\text{ for normal incident and as } T_p = 2Z_i \cos \theta/(Z_2 \cos \theta + Z_1 \cos \theta_i)\text{ in the case of oblique incidence. The angles are related by Snell’s law.} \]
which easily can be derived using the relationship

corresponding equations expressing the intensity transmission are determined by the acoustic impedances of the two media.

\[
T_i = \frac{I_i}{I_t} = \frac{T_r Z_1}{Z_2}, \quad T_r = \frac{4Z_1Z_2}{(Z_2 + Z_1)^2}
\]

\[
T_r = \frac{4Z_1Z_2 \cos^2 \theta_i}{(Z_2 \cos \theta_i + Z_1 \cos \theta_r)^2}
\]

which easily can be derived using the relationship \( I = p^2/2Z \) (see Intensity).

**Related Articles:** Acoustic impedance, Intensity, Snell’s law, Reflection coefficient

**Transmission ionisation chamber**

*(Radiation Protection)* The transmission ionisation chamber is used for verifying the configuration of the output beam of a linear accelerator in external beam radiotherapy, for example for intensity-modulated radiation therapy (IMRT). Transmission ionisation chambers are also used in diagnostic radiology to measure the dose-area product of an x-ray beam to give a measure of the total energy of the x-ray beam incident on the patient, and therefore to provide an indication of the patient dose.

Usually the transmission ionisation chamber is a flat, multi-wire chamber which is fitted at the exit port/collimator of the linear accelerator or x-ray tube.

A transmission ionisation chamber can also be used in conjunction with other detectors in ionising particle identifier systems.

**Related Article:** Ionisation chamber


**Transmit coil**

*(Magnetic Resonance)* The transmit coil in MRI transmits RF energy pulses. The purpose of the transmit coil is to flip magnetisation in the patient in a prescribed manner. The time course and magnitude of flip angles applied is determined by the given sequence in use. The transmit coil is driven by an RF power amplifier. The output of the amplifier consists of RF pulses shaped to give the RF amplitude required to deliver a specific flip angle over a specific bandwidth of resonant frequencies. The transmit coil forms part of a tuned resonant circuit in order to minimise the driving power requirements of the coil.

The transmit coil must produce uniform flip angles throughout the volume of interest. Reciprocity exists between transmit and receive coils, in that a receive coil with good uniformity of sensitivity to signal throughout a volume will act as a transmit coil with an equivalent uniformity of flip angle throughout the same volume. As such, volume coil designs are suitable as RF transmit coils, whereas surface coil designs are not.

In many examinations in typical clinical imaging systems the body coil is used as the transmit coil with the body coil itself or other coils acting as the receiver(s). The bulk of coils used with clinical imaging systems are receive only. A coil that combines transmission and receive function is called a ‘transceiver’ coil.

**Transmitter**

*(Ultrasound)* An ultrasonic transducer acts most often both as a transmitter and as a receiver. It converts electrical energy to acoustical and acoustical energy to electrical. However, in some ultrasound applications, separate transducers are used for transmission and reception. This is the case, for example for continuous wave Doppler devices and ultrasound bone densitometry.

**Related Articles:** Transducer, CW doppler

**Transmutations of elements in radioactive decay**

*(Nuclear Medicine)* The process of generating a new element via decay from an initial element. One way to achieve this is to irradiate a stable isotope with neutrons. Neutrons interact with the nucleus increasing the mass number which yields a new radioactive isotope. The radioactive isotope decays via alpha or beta decay (decreases the atomic number) which yields a new element.

**Related Articles:** Mass number, Atomic number
Transparency

(Diagnostic Radiology) See Opacity

Transport index

(Radiation Protection) The transport index (TI) is a value attributed to radioisotope packages in order to provide control over radiation exposure. The TI for packages, overpacks, freight containers, etc. shall be determined in the following way:

1. The TI is the maximum radiation level in millisievert per hour (mSv/h) at distance 1 m from the external surface, multiplied by 100. For uranium and thorium ores and their concentrates, the maximum radiation level at 1 m from the external surface of the load might be taken as: 0.4 mSv/h for ores or physical concentrates of uranium and thorium; 0.3 mSv/h for chemical concentrates of thorium; 0.02 mSv/h for chemical concentrates of uranium (except uranium hexafluoride).

2. For tanks, freight containers, etc., the aforementioned TI value shall be multiplied by appropriate factors depending on the dimension of the transport. Taking into account the surface, in square meters, of the biggest surface of the package, the multiplicative factors are: 1 for a surface of 1 m²; 2 for a surface between 1 and 5; 3 for a surface between 5 m² and 20 m²; 10 for a surface of more than 20 m².

The values of TI obtained earlier shall be always rounded up to the first decimal place. Values of 0.05 or less may be considered as zero.

The transport index of overpacks, freight containers, and conveyance shall be the sum of the TIs of each package contained, or shall be determined by direct measurement of radiation level. In case of non-rigid overpacks the transport index shall be the sum of the TIs of all packages.

Transrectal transducers

(Ultrasound) Transrectal ultrasound is used for examining the prostate. Imaging sections of the prostate should be ideally obtained in two planes for diagnostic: true transverse axial plane and sagittal plane. For rectal ultrasound-guided biopsy, an end viewing sagittal section is needed and for perineal biopsy a sagittal plane view is best (Richards and Kelly, 1996).

Three types of transducers are used, depending on the application, for transrectal examination: Single plane probes, with a linear array of elements; biplane probes with one linear and one curved array set at a right angle to each other; and multi-plane probes, with a single transducer orientated mechanically or as a phased array (McDicken, 1996; Richards and Kelly, 1996).

Most transrectal ultrasound transducers either have or can be fitted with a biopsy guide and needle.


Transversal magnetisation

(Magnetic Resonance) Transversal magnetisation, otherwise known as transverse magnetisation, refers to magnetisation lying in the transverse plane, defined as the plane perpendicular to the direction of the static magnetic field. It is changing transversal magnetisation that leads to the generation of an NMR signal and, hence, ultimately to the formation of MR images.

When an object is placed in a static magnetic field, nuclei with spin \( I = \frac{1}{2} \) occupy one of two energy levels, corresponding to orientation of the \( z \)-component of angular momentum parallel (lower energy) or antiparallel (higher energy) to the static field direction. In this steady state condition, the greater population of the lower energy level (determined by the Boltzmann distribution) generates a net longitudinal magnetisation along the direction of the field. Random precession of spins about the static field (or, in a quantum mechanical model, Heisenberg’s uncertainty principle applied to angular momentum) ensures that there is no coherent magnetisation in the transverse plane.

This situation is altered by the application of a radiofrequency (RF) magnetic field at the Larmor frequency of the spins. This results in promotion of spins from the lower to the higher energy level and brings these spins into phase coherence. Thus the longitudinal magnetisation is decreased in size and a net transverse magnetisation is generated. The extent of these phenomena depends on the amplitude and duration of the RF field, commonly expressed as the flip angle or rotation angle. Following a 90° pulse, longitudinal magnetisation is completely eliminated and all of the magnetisation lies in the transverse plane.

Transverse magnetisation is composed of the individual magnetic moments of the coherent spins, and hence precesses about the static field direction at the Larmor frequency. It also decays due to spin–spin relaxation. This changing magnetisation induces an electric current in a neighbouring receive coil according to Faraday’s law of induction, and it is this current that constitutes the NMR signal.

An NMR signal detected immediately after excitation is known as a free induction decay (FID) signal. More commonly, however, signal acquisition is delayed using gradient echo and/or spin echo techniques, both of which involve manipulation of transverse magnetisation to generate a signal later in time. The resulting echo signal is \( T_2^* \) or weighted, depending on the technique employed.

MRI data acquisition almost invariably requires repeated acquisition of a pulse sequence with excitation repeated at an interval known as the repetition time (TR). Usually TR is long enough that complete \( T_2^* \) decay of transversal magnetisation may be assumed to have occurred before application of the next excitation pulse. However, in some instances, particularly fast gradient echo sequences, this may not be the case, and it is then necessary to consider the effect of excitation pulses on transversal magnetisation as well as longitudinal magnetisation, leading to much more complicated spin evolution and contrast properties.

Related Articles: Boltzmann distribution, Excitation, Flip angle, Free induction decay (FID), Gradient echo (GE), Longitudinal magnetisation, Receive(r) coil, Repetition time (TR), Spin echo, Spin–spin relaxation, \( T_2^* \)-weighted, \( T_2^* \).

Transversal plane

(General) See Anatomical body planes

Transverse wave

(Ultrasound) When an ultrasound wave propagates in a medium, the particles within the medium will start oscillating. In gases, liquids and soft tissue this oscillation is always in the same direction as the ultrasound wave, giving rise to longitudinal wave propagation. However, in solid materials both longitudinal (compressional) and transverse (shear) wave propagation are possible. In a transverse wave the particles oscillate perpendicular to the wave propagation direction. Figure T.48. This is due to the stronger bonds between atoms and molecules in solid materials compared to the much weaker bonds in liquids and gases.
When a longitudinal wave crosses the boundary (at an angle) between two solid materials, two different reflected waves and two different transmitted waves would occur (Figure T.49): one longitudinal wave and one transverse wave. This is often called mode conversion. The transverse waves propagate at a slower speed of sound than the longitudinal, giving rise to the different angles (by Snell’s law) of the transmitted and reflected waves. This phenomenon is exploited in ultrasound-based non-destructive material testing.

**Related Articles:** Longitudinal wave, Surface wave, Lamb wave

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The target used to create a MV photon beam is normally made from Tungsten, and is placed in the path of the electron beam. The Bremsstrahlung photons produced by the interactions in the target will be highly peaked along the central axis. Hence a flattening
conical filter is used to reduce the intensity in the centre. This filter is energy specific.

The range of targets and filters are mounted on a rotating carousel or a sliding mechanism to enable automatic positioning within the beam.

The treatment head also contains the primary pinhole collimator, which defines a maximum circular field, and secondary adjustable rectangular collimators, which can be moved independently to create a rectangular field. The leakage through these collimators should be checked regularly as part of a quality control program. Modern linacs may also incorporate multileaf collimator blocks within the treatment head; however, these can also be externally added.

If an electron beam is required, the photon target is moved out of the path of the electron beam, and is replaced by a thin beryllium window and two scattering foils. These modify the shape and spectrum of the intense pencil beam of electrons to produce a clinically useful beam. Electron beams require additional collimation called applicators to reduce the dose from the high scatter that occurs in air. The beam current in electron mode must be reduced by a factor of 100 (along with the frequency of the RF source) from that of x-ray mode to prevent dangerously high dose rates for electrons.

**Cobalt Unit:** The treatment head of a cobalt unit consists of:

- Source housing (the source head)
- Collimators
- Light field to show the extent of the radiation field when the source is in the OFF position

The source head consists of shielding whilst the source is in the OFF position (steel shell with lead), and a mechanism for moving the source from OFF to the ON position by the collimator opening. This can either be achieved by moving the source to the treatment position, or moving attenuation material in front of the source. Three methods are as follows:

1. A rotating shutter to expose or shield the source
2. A rotating heavy metal drum in which the source is mounted
3. A sliding source bar

**Related Articles:** Linear accelerator, Multileaf collimators, Cobalt unit

### Treatment optimisation

*(Treatment Planning)* Treatment optimisation is optimisation of the radiotherapy treatment plan, particularly in external beam x-ray planning. For each treatment beam, an optimisation algorithm is used to generate the beam profile needed to deliver the prescribed dose distribution using inverse planning. After the plan has been optimised a second stage is often involved in which the constraints of the delivery system are modelled to produce a deliverable beam plan. This is synonymous with fluence optimisation.

**Related Articles:** Fluence optimisation, Interactive planning, Inverse radiotherapy planning, Simulated annealing algorithm

### Treatment parameters

*(Radiotherapy)* Treatment parameters determine the shape and the size of treatment field, the use of wedges or any device for beam modification, angles of beams and number of monitor units used to deliver the required radiation dose. They will vary for individual patients and contain beam data including number and position of treatment beams relative to the patient’s contour and size, tumour size and location and the external reference points.

The treatment parameters are included in the radiotherapy treatment chart which is an important working instrument, not only as a set of data but also as a method of communication among radiation oncologists, physicists and technicians. In addition to administrative and medical data, physical and simulation data that are indispensable for the daily accurate reproduction of the therapy procedures should be recorded as well as accurate daily entries of the fractional and cumulative absorbed doses. Any quality assurance programme must rely on the accessibility of the radiation treatment history and a correct record of the therapy protocol in order to be verifiable.

### Treatment phase

*(Radiotherapy)* Often radiotherapy treatments are delivered in a set of stages often referred to as phases. The use of several phases is often done to enable a sequential boost of the dose to the region closest to the centre of the target volume. Examples include the use of a boost phase in breast treatment to give more dosage to the tumour than to the surrounding region.

**Related Articles:** Radiotherapy treatment planning, Boost dose

### Treatment plan evaluation

*(Radiotherapy)* In radiotherapy a treatment plan is often evaluated in comparison with other competing plans to determine which is the best plan with which to treat the patient. This evaluation may be visual inspection of the isodose coverage of the plan on the screen of the TPS or in a printout, or it may be a more quantitative evaluation based on dose–volume histograms (DVHs) or calculation of predicted tumour control probability (TCP) or normal tissue complication probability (NTCP).

**Abbreviations:** DVH = Dose–volume histogram, NTCP = normal tissue complication probability, TCP = Tumour control probability and TPS = Treatment planning system.

**Related Articles:** Dose area histogram, Dose–volume histogram, Dose–volume histogram differential (DVH), Dose–volume histogram integral (cumulative) DVH, Probability of complications, Tumour control probability (TCP), Normal tissue complication probability (NTCP)

### Treatment plan normalisation

*(Radiotherapy)* The process of normalising the treatment plan dose distribution to achieve a result consistent with dose–volume constraints, on the target and organs at risk.

**Related Article:** Target dose

### Treatment planning system

*(Radiotherapy)* The treatment planning system consists of a number of computer workstations networked together and also with other equipment within the hospital (e.g. simulators, CT and MR scanners, and linacs through a record and verify system, etc.). Associated with the workstations will be high resolution, large monitors for displaying images and graphics packages for visualising information in three dimensions. This has allowed the user to view the dose coverage of tumours as a volume rather than on just one CT slice, to display the beam’s eye view of the treatment field, to calculate DVHs, and the ability to generate DRRs. The increased power and speed of computers has allowed complex dose calculations to be performed very rapidly and now even more complex calculations (e.g. IMRT and electron Monte Carlo) are possible in a reasonable time. Other equipment associated with the treatment planning computer includes plotters, printers, film scanners and digitisers. The computer must also have sufficient memory for storage of information, and this is becoming more relevant with the rapid increase in additional imaging being performed. It is also essential that safety is built into the
system and it is robust to prevent problems such as viruses, power failures, etc. There should be a mechanism for a regular back-up or archiving of the information stored on the system, and also regular QA tests performed on the planning system.

Before using a planning system the user must input mechanical specifications and limits, naming conventions, and dosimetric data relating to the linacs used, which can then be used as required when calculating a plan. There are a number of different calculation algorithms which can be used for calculating the treatment plan. Further details on this are given in the following references.

**Abbreviations:** CT = Computed tomography, DRR = Digitally reconstructed radiograph, DVH = Dose–volume histogram, IMRT = Intensity-modulated radiation therapy, MR = Magnetic resonance and QA = Quality assurance.

**Related Articles:** Kernel-based treatment planning, Pencil beam


### Treatment planning systems (brachytherapy)

*(Radiotherapy, Brachytherapy)* A modern treatment planning system for image-guided brachytherapy (BT) includes the same type of basic modules and tools as an external beam radiotherapy (EBRT) treatment planning system for many planning tasks. The first tasks/modules encountered in the treatment planning process are the modules for importing and registering 3D image volumes and the modules for definition of tumour and target volumes and organs at risk. Both EBRT and BT systems need tools to display the 3D dose distributions in relation to anatomy, together with evaluation tools, for example based on dose–volume histograms. Algorithms for dose distribution optimisation are also needed, even though they might be of different types for EBRT and BT. Facilities for documentation of approved plans are also required, as well as access to appropriate archiving, information and communication systems.

**Treatment Planning Systems – Brachytherapy:** Specific for a brachytherapy treatment planning system are modules for definition of applicators and sources (source stop positions). For image-guided brachytherapy the system should allow accurate interactive definition of applicators and source stop (dwell) positions in the 3D volume. Radiograph-based applicator/source definition is still used, and so this technique must also still be supported (use of digitisers, import of film). The availability of a standard applicator library of ‘rigid’ applicators in the planning system greatly assists the process.

Another brachytherapy specific module is the dose calculation module, requiring a source model. Source models have been based on point sources and line sources, and a popular source model is the model based on the AAPM TG43 formalism. In contrast to EBRT, where acceptance testing and commissioning of a treatment planning system requires a number of measurements for the treatment units to be included, in BT this procedure consists of a verification process. The important task of the physicist is not to make measurements around the brachytherapy source but to verify, first, which source model is used in the system, and second, that this source model is implemented correctly for the source used, where both the isotope and the source design must be correct. Verification is performed by comparing treatment planning system calculations with recognised published ‘best source data’. These data are available from the AAPM and from ESTRO.

Brachytherapy dose distributions are characterised by large dose variations inside the target volume. Besides the traditional cumulative and differential dose–volume histograms, the natural dose–volume histogram is also used for evaluation purposes, where a point source is characterised by a horizontal line. As already mentioned, some dose optimisation algorithms and tools are specific to brachytherapy. Some tools are very useful but must be used with care. An example is the ‘grab and drag’ tool which allows the user to move isodose lines and when used in this way has a profound effect on dwell times at the defined stop positions in the applicators.

For specific brachytherapy applications, dedicated treatment planning systems are available. One such application is ultrasound-guided interstitial permanent implantation of seeds for prostate cancer. The dedicated systems are streamlined for the seed implantation process, and allow the user, for example to follow the implantation procedure interactively and continuously update the seed positions and the resulting dose distribution.

**Treatment Planning Systems – Brachytherapy, Future Developments:** Traditionally, brachytherapy planning was based on ‘dosimetry systems and orthogonal radiographs’ without actual definitions of volumes of interest. Dose calculations were based on point dose models in an infinite medium, and in principle no corrections were made for heterogeneities.

With the development of image-based brachytherapy there are now available 3D definitions of volumes of interest, information on tissue densities and on applicator/source positions. We also have afterloading units, low energy sources, Monte Carlo calculations, algorithms for the optimisation of physical dose and for radiobiological effects, etc. and there is a renewed interest in the development of brachytherapy source models and dose calculations.

The treatment planning systems using the TG43 formalism still have a number of limitations:

- There are no heterogeneity corrections (water medium).
- There are no corrections for patient size (‘infinite’ medium).
- There are in general no corrections for applicator composition and possible shields.
- There are no intersource effects included.
- There are no transit dose corrections for remotely controlled afterloading units.

In general, new algorithms based on Monte Carlo calculations are needed.

In the clinical setting, where you work with one or several brachytherapy treatment planning systems, get to know your systems:

- Understand the source model/s used.
- Understand how source strength is defined.
- Understand the way source decay is handled.
- Understand how source positions are determined in the system.
- Understand how the source actually moves in the applicator during treatment, in order to treat according to the plan.
- Get to know your applicators.
- Understand how the system is ‘constructed’.
- Understand the limitations of the system.
- ‘Beware of updates!’ They might change more than they are supposed to change! ‘Improvements’ might not be improvements for you!
**Abbreviations:** AAPM = American association of physicists in medicine and ESTRO = European society for therapeutic radiology and oncology.

**Related Articles:** Source models, Point source calculation, AAPM TG43 formalism, Dose–volume histograms – brachytherapy, Orthogonal films, Interactive implant technique, Dwell times, Intersource shielding


**Treatment position**
(Radiotherapy) This is the position in which the patient is treated, and is determined by practical constraints. It needs to be highly reproducible, and such that it is possible to treat the target. This is achieved by immobilisation devices and verification techniques such as image-guided radiotherapy (IGRT).

**Related Articles:** Immobilisation, IGRT

**Treatment room**
(Radiotherapy) This is the location where the patient will receive their radiotherapy treatment; it is also commonly known as a bunker. The design of the treatment room must be such that the level of dose outside the room is below that required by legislation. The room design will be dictated by the type of treatment machine and the energy of the treatment beam. Once a machine is capable of producing a beam following installation, it is essential that a full radiation protection survey be carried out to confirm the integrity of the design.

In the case of a superficial unit (operating in the kV region, e.g. approximately 100 kVp), some additional protection such as lead lining around the door and leaded glass will be necessary. The leaded glass allows the operator to directly observe the patient during treatment.

For higher energy units in the MV region of operation (e.g. linear accelerators, cobalt units), a much higher level of protection is required. These rooms will either make use of a maze entrance or a heavy lead-lined door (in situations where space is at a premium) to ensure that the scattered dose at the entrance to the room is low enough to meet the requirements. The room walls will consist of thicker portions of concrete (usually around 2.5 m thickness) known as the primary barrier where the beam can directly impinge onto the walls, ceiling and floor. Thinner portions of concrete (typically around 1.5 m thickness) known as the secondary barrier are used elsewhere where the beam cannot reach directly. When planning the thicknesses of material required, it is essential to be aware of occupancy in the surrounding areas, as well as above and below the treatment room.

When linear accelerators with x-ray beams of energy greater than 10 MV are to be employed, the production of neutrons must be accounted for in the shielding design of the room.

In order for the treatment unit operator to observe the patient during treatment with an MV treatment unit, a CCTV system is put in place. Other additional features of treatment rooms are door interlocks (to turn the beam off immediately if someone attempts to enter the room while the beam is on), air venting systems as well as the capability to allow full movements of the treatment machine, storage space for the associated accessories and sufficient space for specialised treatments.

**Related Articles:** Interlock; Interlocking device, Isocentre, Leakage radiation, LINAC, Linear accelerator, Cobalt unit, Maze, Kilovoltage (kV), Orthovoltage, Secondary barrier, Use factor, Total body irradiation.

**Trendelenburg position**
(General) There are a series of terms used to describe the position of an individual when undertaking different imaging examinations. ‘Trendelenburg’ means tipping the individual head down.

**Related Article:** Patient position

**Trigger delay**
(Magnetic Resonance) In an ECG gated cardiac MRI, the trigger delay TD is the time between detection of an R–wave trigger and the start of the acquisition of lines in k-space. A TD can be used to ensure data acquisition occurs during diastole, when motion artefact will be reduced.
Triggering

(General) In general, a ‘trigger’ is any event that initiates some action. In electronics it refers to any electrical signal which causes some event to occur in a synchronised manner.

It is common when some function is required to occur at a fixed time relative to another event, to derive some electrical signal from this first event, and use it to ‘trigger’ the required function. It is particularly useful when multiple repeat events need to be synchronised to external stimuli (e.g. x-ray exposures synchronised to the ECG).

In a cathode ray oscilloscope (CRO) a ‘trigger circuit’ is provided, along with a variable ‘trigger level’ control which selects a particular input level or threshold for the incoming signal to ‘trigger’ the trace to cross the screen. In this way the traces on the CRT screen can be made to appear as one stationary trace (Figure T.51).

Related Article: Threshold detection

Triode tube

(Diagnostic Radiology) The triode tube is a highly evacuated electron tube containing an anode, a cathode and a control grid.

The principle of its operation is that, as with a thermionic diode, the heated filament causes a flow of electrons that are attracted to the plate and create a current flow. Applying a negative charge to the control grid will tend to repel some of the (also negatively charged) electrons back towards the filament: the larger the charge on the grid, the smaller the current flow to the plate. If an AC signal is superimposed on the DC bias of the grid, an amplified version of the AC signal appears in the plate circuit.

Triode tubes are the predecessor of transistors and are used in the control circuits of some high voltage equipment (Figure T.52).


True coincidences

(Nuclear Medicine) True coincidences are used in PET imaging and refer to registered coincidences where the two 511 keV photons originate from the same annihilation. The opposite of true coincidences are false coincidences, that is where the two photons generating a coincidence are either not from the same annihilation or one or both photons are scattered before registration.

Abbreviation: PET = Positron emission computed tomography.

Related Article: Event type in PET

True negative

(General) The diagnosis of a disease using some kind of modality, such as a scintillation camera, involves some kind of uncertainty in the decision of whether a patient has an abnormality or is normal. There can be four alternatives:

1. The diagnosis indicates a positive answer (disease) to a patient that has a disease.
2. The diagnosis indicates a positive answer (disease) to a patient that does not have a disease.
3. The diagnosis indicates a negative answer (no disease) to a patient that has a disease.
4. The diagnosis indicates a negative answer (no disease) to a patient that does not have a disease.

The earlier situations are often labelled as follows:

1 = true positive (TP)
2 = false positive (FP)
3 = false negative (FN)
4 = true negative (TN)

The fraction of true positive to all cases is called the sensitivity, TP/(TP+FN), and the fraction true negative to all cases is called specificity TN/(TN+FP). A receiver-operator characteristic (ROC) analysis determines the fractions of 1–4. The result is usually plotted as a ROC curve in the following Figure T.53 shows such a curve of two imaging systems where system B is better than system A.

Related Articles: True positive, False positive, False negative, ROC

True positive

(General) See True negative

Tube current (mA)

(Diagnostic Radiology) The thermal electrons, accelerated towards the anode by the electric field between the cathode
and anode, form the anode current \( I_a \) (also known as x-ray tube current – \( I_a \)). In practice it is expressed in mA, as it normally is within this range of current. The tube current is proportional to the production of thermal electrons (filament heating). At lower kV (usually below 50 kV) the tube current cannot increase with further increase of the filament temperature (i.e. generation of more thermal electrons). This is due to the Space charge effect, and in this case the tube current can be increased only by increasing the kV (see article on filament current).

Tube current does not influence the contrast of the x-ray image (but changes its brightness). The operating mA depends on the attenuation of anatomical region to be imaged. Due to this reason the radiographic tube current could vary significantly (e.g. from 5 to 1000 mA). The fluoroscopic tube current is usually very small (e.g. from 1 to 4 mA), due to the high intensifying factor of the image intensifier.

The intensity of x-ray radiation \( W \) (x-ray energy flux density) is \( W \sim I U^2 Z \), where \( I \) is the anode current \( (I_a) \) \( Z \) is the atomic number of anode \( U \) is the accelerating high voltage \( (U_a) \).

From this is seen that the intensity of the x-ray radiation (hence the exposure) is in linear relation with the mA.

The tube current is usually measured with mA-meter placed between ground and the middle point of the high voltage transformer (see the article on high-voltage generator and its circuits).

The variation of the \( I_a \) with changes of \( U_a \) (using the filament current \( I_f \) as a parameter) is known as the anode characteristic of the x-ray tube (Figure T.54).

**Related Articles:** High voltage generator, High voltage circuit, Filament circuit, Filament current, Stationary anode, Rotating anode, Target


**Tube filament current**
(Diagnostic Radiology) See Filament current

**Tube housing**
(Diagnostic Radiology) See X-ray tube housing

**Tube load**
(Diagnostic Radiology) Tube load is a measure representing the energy imparted to the x-ray tube anode (kV mA). Approximately

**Tube kilovoltage (kV)**
(Diagnostic Radiology) The accelerating voltage between the anode and cathode of an x-ray tube is also known as tube kilovoltage (or anode voltage). In practice it is expressed as the peak kV kVp (in practice it is often referred to as just kV). Varying the tube kilovoltage varies the energy (keV) of the x-ray photons produced by the x-ray tube, thus forming the x-ray spectrum. The keV vary from 0 to the applied tube kilovoltage (e.g. if \( U_a = 60 \), than photons energy vary from 0 to 60 keV).

Tube kilovoltage determines the contrast of the x-ray image. Due to this reason the operating kV depends on the anatomical region to be imaged. The radiographic and fluoroscopic tube kilovoltage is usually between 50 and 150 kV. Mammographic tube kilovoltage is between 20 and 35 kVp.

The intensity of x-ray radiation \( W \) (x-ray energy flux density) is \( W \sim I U^2 Z \), where \( I \) is the anode current \( (I_a) \) \( Z \) is the atomic number of anode \( U \) is the accelerating high voltage \( (U_a) \).

From this is seen that the intensity of the x-ray radiation (hence the exposure) is in quadratic relation with the kV. Similarly the \( U_a \) (kV) influences far more the contrast than the \( I_a \) (mA).

The tube kilovoltage is usually measured indirectly at the primary (low voltage) circuit of the high voltage transformer (based on the fact that this transformer has fixed transformation ratio). Another method to measure the kilovoltage indirectly is by the kVp meter – effectively measuring the maximum photon energy of the x-ray beam (peak kV, or kVp).

The variation of the \( I_a \) with the change of \( U_a \) (kVp), using the filament current \( I_f \) as a parameter, is known as the Anode characteristic of the x-ray tube (Figure T.54).

**Related Articles:** Tube current, High voltage generator, High voltage circuit, Filament circuit, Target


**Panel T.53** ROC curves comparing two imaging systems (A and B).

**Panel T.54** The graph shows the variation trend of the tube current \( I_a \) (mA) with changes of the anode voltage \( U_a \) (kV) at three different \( I_f \) (three different increasing temperatures of the cathode filament).
99% of this energy transforms to heat (strictly speaking the energy converted directly to heat is ~75%); hence, the maximum permissible tube load depends on the heat capacity of the anode heat units (HU).

Tube load depends on the time (length) of the exposure – a single short exposure can impart much more energy to the anode, as after it will follow a long cooling period (during the time when the current patient leaves and a new patient arrives). Tube load for long exposures depends very much on the cooling effectiveness of the x-ray tube (e.g. fast rotating anode using oil cooling with external heat exchanger cools quickly the anode during the exposure, hence allows more energy to be imparted in it).

Various x-ray tube operating charts can be used to determine the tube load at specific kV, mA and ms. Some high voltage generators include a special calculator which estimates the tube load after each exposure and controls the cooling process. This automatic calculator rejects parameters which will overload the x-ray tube, and also ceases the current exposure(s) if the maximum temperature is exceeded (this way preventing the anode from thermal damage).

A specific operating mode of the x-ray equipment (falling load mode) is sometimes used to allow the most effective use of the heat capacity of the anode and this way allows production of very powerful exposures for very short time.

**Related Articles:** Heat units, Heat capacity, Maximum load, Falling load, Cooling curve, Tube rating charts

**Tube load time**

(Diagnostic Radiology) Tube load time is a term associated with the time of the x-ray exposure. Sometimes the term can be used to specify the time for which an exposure reaches the peak power of the requested exposure (the front porch of the x-ray exposure pulse). Also sometimes the term can be used to specify the maximal permissible time of the x-ray exposure (for which the tube reaches its maximum load).

**Related Articles:** Heat units, Heat capacity, Maximum load, Falling load, Cooling curve, Tube rating charts

**Tube rating charts**

(Diagnostic Radiology) Normally the manufacturers combine the heat loading characteristics and limitations in diagrams known as tube rating charts, anode load charts, or other. These charts depend not only from the x-ray tube parameters, but also from the rectification and other specifics of the high voltage generator. The charts are given according to the focal spot size, speed of anode rotation, kW waveform, etc. These charts are also given for different modes of operation: single exposure, multiple exposures, cine-radiographic exposures, etc.

Contemporary x-ray equipment have all these charts built in special automatic calculators. This way the equipment automatically rejects parameters, which will overload the x-ray tube (and sometimes suggests other parameters). The design of this automatic varies widely – from electronic analogous systems to microprocessor programs.

**Figure T.55** represents an example for a single exposure rating chart for low power rotating anode x-ray tube. The curves (limiting curves) represent the maximum permissible tube current (Ia) for various length of exposures (ms).

If such a graph is given for multiple exposures (e.g. when sequence of exposures is used in angiography), the chart would represent the maximum permissible heat unit as a function of the total number of exposures. In this case, the parameter will be the number of exposures per second. Similarly, if the chart is for cine use (again in cardiac angiography), then the maximum permissible heat units will be plotted against the number of cine frames (in this case the parameters for the limiting curves will be the length of one cine frame).

**Examples from Figure T.55**

- Exposure with 100 mA and 50 ms is acceptable for all kVs, as it is below all limiting curves (i.e. the cross point of 100 mA and 50 ms is below the all limiting curves).
- Exposure with 400 mA and 20 ms is not acceptable with 100 kVp, as it will impart too much energy to this particular anode, but the same exposure is acceptable with 80 kVp.
- Exposure with 500 mA and 30 ms is not acceptable even with 80 kVp, but if the speed of anode rotation increases from the normal 3000 to 9000 rpm, then the tube cooling is better and the exposure is acceptable (i.e. the cross point of 400 mA and 100 ms is above 80 kVp curve, but is below the dotted limiting curve for 80 kVp at 9000 rpm).
- Exposure with 800 mA and 500 ms will not be permitted with any kVs, as it will overheat the anode, hence destroy the x-ray tube.

**Related Articles:** Heat units, Heat capacity, Maximum load, Falling load, Cooling curve, Tube rating charts

**Tube stand**

(Diagnostic Radiology) Tube stand and tube crane are mechanical supports which hold the x-ray tube housing and its light beam diaphragm (with collimator). These supports can be floor or ceiling mounted.

**Tumour antivasculary alpha therapy**

(Background) Radioisotopes are used in nuclear medicine procedures for imaging and therapy. Imaging isotopes emit gamma rays; therapeutic isotopes emit low energy gamma rays or high
energy beta radiation. Cancer is the main target for therapeutic nuclear medicine. A new approach to therapy is emerging where radioisotopes that emit very short range (80µm) alpha particles are tagged onto monoclonal antibodies for targeted alpha therapy (TAT). The alpha radiation is high linear energy transfer (LET) radiation and transfers α particles to the targeted cells, causing increased double strand breaks in the nuclei of the targeted cancer cells. The radiation weighting factor for alphas is 20 and the relative biological effectiveness (RBE) for tumour regression is approximately 3–5.

TAT was not indicated for solid tumour therapy, as the range of alphas is too short, the diffusion time into the tumour too long and the uptake too heterogeneous.

Tumour anti-vascular alpha therapy (TAVAT) overcomes all these obstacles. The AIC can diffuse into the peri-vascular space through tumour capillary fenestrations, and target antigens expressed by pericytes and adjacent cancer cells. These antigens will filter out the AIC and, on decay of the alpha radioisotope, will set up a longitudinal alpha field, causing DSBs in the endothelial cell nuclei, apoptosis and closure of the tumour capillary. Oxygen and nutrient starvation will lead to tumour regression if enough capillaries are shut down.

Only alpha radiation can achieve this effect, the indirect killing of tumour capillary endothelial cells, and regression of tumours by antivascular therapy.

**Abbreviations:**
- AIC = Alpha-immunoconjugate
- DSB = Double strand break
- LET = Linear energy transfer
- MTD = Maximum tolerance dose
- RBE = Relative biological effectiveness
- TAT = Tumour alpha therapy
- TAVAT = Tumour antivascular alpha therapy

**Related Article:** Targeted alpha therapy

**Tumour control**

(Radiotherapy) The aim of radiotherapy is to affect tumour control without inducing serious complications due to the irradiation of normal tissue. For tumour cure it is generally accepted that all clonogenic cells must have suffered reproductive death and so be incapable of further multiplication. For more details see the article on tumour control probability.

**Related Article:** Tumour control probability

**Tumour control probability (TCP)**

(Radiotherapy) The relationship between dose and tumour control probability (TCP) is shown in Figure T.56. It has a sigmoid (S) shape with the probability of tumour control tending to zero as the dose tends to zero and tending to 100% at very large doses.

The sigmoid shape can be explained from the random nature of cell killing after irradiation and the need for all clonogenic cells to have suffered reproductive death and so be incapable of further multiplication. There are those that believe tumour cure can be obtained by sterilising most of the clonogenic cells rather than all of them as the body’s immune system will destroy those remaining but there is no real evidence to support this.

Taking the theory that radiation sterilisation of all clonogens is required, the probability of tumour control as a function of the radiation dose can be derived by the application of Poisson statistics to cell survival curves. Therefore, the probability of tumour control may be described as in Equation T.8 where SF is the surviving fraction, obtained using the linear quadratic model, and N∞ is the original number of tumour clonogens in the target volume:

\[
TCP = e^{-SF N_{\infty}}
\]

An expression for tumour control probability can be obtained by the application of Poisson statistics to cell survival curves.

It should be noted that this model describes the response of individual tumours and the dose–response curves obtained from it have particularly steep slopes. In reality, the position and steepness of the dose–response curve will vary within a tumour cell population due to their heterogeneous nature and will depend on the volume of tissue irradiated (dose–volume effect), the fractionation scheme and if the treatment includes concurrent chemotherapy. However, the Poisson method can be adapted to incorporate such effects and many authors have suggested such models although there is not yet a single, established model in widespread use. A comprehensive review of TCP models can be found in Chapter 10 of the book by Dale and Jones (2007).

In clinical practice, the achievable level of tumour control will depend on the tolerance of neighbouring normal tissues. For more information see the article on Therapeutic effect.

**Abbreviation:**
- TCP = Tumour control probability

**Related Articles:**
- Cell survival curves, Dose–response model
- Linear quadratic (LQ) model, Radiosensitisers, Sigmoid dose–response curve, Surviving fraction, Therapeutic effect, Tolerance

**Further Readings:**

**Tungsten (General)**

<table>
<thead>
<tr>
<th>Symbol</th>
<th>W</th>
</tr>
</thead>
<tbody>
<tr>
<td>Element category</td>
<td>Transition metal</td>
</tr>
<tr>
<td>Mass number A of stable isotopes</td>
<td>180 (0.12%), 182 (26.50%), 183 (14.31%), 184 (30.64%), 186 (26.43%)</td>
</tr>
<tr>
<td>Atomic number Z</td>
<td>74</td>
</tr>
<tr>
<td>Atomic weight</td>
<td>183.84</td>
</tr>
<tr>
<td>Electronic configuration</td>
<td>1s2 2s2 2p1 3s2 3p6 3d10 4s2 4p6 4d10 5s2 5p6 4f14 5d6 6s2</td>
</tr>
<tr>
<td>Melting point</td>
<td>3695 K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>5828 K</td>
</tr>
<tr>
<td>Density near room temperature</td>
<td>19.25 × 10³ kg/m³ (19.25 g/cm³)</td>
</tr>
</tbody>
</table>

**History:** Tungsten (aka Wolfram) was first isolated in 1783 by the brothers José and Fausto Elhuyar, who used charcoal to reduce tungstic acid.
**Isotopes of Tungsten:** There are five isotopes of tungsten that are considered stable, though they are in fact metastable isotopes with extremely long half lives. These include the following:

<table>
<thead>
<tr>
<th>Isotope of tungsten</th>
<th>Half life</th>
<th>Natural abundance</th>
</tr>
</thead>
<tbody>
<tr>
<td>180W</td>
<td>1.8 ± 0.2 × 10^9 years</td>
<td>0.12%</td>
</tr>
<tr>
<td>181W</td>
<td>&gt;8.3 × 10^9 years</td>
<td>26.50%</td>
</tr>
<tr>
<td>182W</td>
<td>&gt;2.9 × 10^9 years</td>
<td>14.31%</td>
</tr>
<tr>
<td>183W</td>
<td>&gt;1.3 × 10^9 years</td>
<td>30.64%</td>
</tr>
<tr>
<td>184W</td>
<td>&gt;2.7 × 10^9 years</td>
<td>28.45%</td>
</tr>
</tbody>
</table>

**Medical Applications:** X-ray tube filament and target – Tungsten compounds and alloys are widely used as both the electron-producing filament and the x-ray producing target in a standard x-ray tube. The high atomic number of tungsten makes it a more efficient generator of x-ray radiation than many other materials (although the bremsstrahlung process is still extremely inefficient) due to greater electrostatic forces between the tungsten atoms and the incident electrons. When tungsten is used as a filament, electrons are released by thermionic emission when a current is passed through the filament.

Shielding of PET radionuclides – The high atomic number of tungsten allows it to be used to shield the high energy radiation produced by some PET radiopharmaceuticals, for example fluorodeoxyglucose (FDG). The tungsten provides a large interaction cross section, enabling significant attenuation of the radiation passing through it.

Scintillation detectors – Tungsten compounds in the form of tungstate crystals are used as the scintillation material in some nuclear medicine imaging devices such as gamma cameras.

**Related Articles:** Fluorodeoxyglucose (FDG), Target of x-ray tube, Radiation shielding, Scintillation camera

**Turbo factor**

(Magnetic Resonance) See Echo train length

**Turbo gradient spin echo (TGSE)**

(Magnetic Resonance) Turbo gradient spin echo (TGSE) is a hybrid pulse sequence that combines gradient echo and spin echo. The sequence is primarily used to obtain $T_2$ weighted images. Compared to fast spin echo (FSE) TGSE has shorter acquisition time, decreased RF power deposition and more sensitivity to magnetic susceptibility differences allowing improved visualisation of haemorrhage. TGSE is also called GRASE (gradient and spin echo). See GRASE for a more detailed description of the pulse sequence.

**Related Articles:** Echo planar imaging (EPI), Fast spin echo (FSE), Gradient and spin echo (GRASE)

**Turbo spin echo**

(Magnetic Resonance) Turbo spin echo (TSE) is a pulse sequence characterized by a 90° pulse followed by a series of rapidly applied 180° rephasing pulses to form a series of spin echoes. Other names for this type of sequence are fast spin echo (FSE), rapid acquisition relaxation enhancement (RARE) and half-Fourier single shot turbo spin echo (HASTE). See Fast spin echo (FSE) for a more detailed description.

**Related Articles:** Echo train length, Fast spin echo (FSE), Half acquisition single-shot turbo spin echo (HASTE), Rapid acquisition relaxation enhancement (RARE)

**Turbulence**

(Ultrasound) Turbulence describes a state of turbulent flow.

(See Turbulent flow.)

**Related Article:** Turbulent flow

**Turbulent flow**

(Ultrasound) Turbulent fluid flow is characterised by random irregular motion. In tubes, although the net flow is in the direction of the tube, turbulence leads to local velocity vectors in several directions with fast changes of velocity in space and time. There may also be non-random features such as vortices and eddies. Turbulence can result spontaneously if Reynolds number exceeds transition values and appears in the human circulation in arterial stenoses. The energy required for turbulent flow is greater than for laminar flow, in arteries this results in increased pressure loss through the turbulent flow (Figure T.57).

**Related Articles:** Laminar flow, Reynolds’ number

**TVL (tenth value layer)**

(Nuclear Medicine) See Tenth value layer (TVL)

**Two-day protocol**

(Nuclear Medicine) The two-day stress/rest protocol is a procedure for myocardial imaging starting with an $^{99m}$Tc-sestamibi or $^{99m}$Tc-tetrofosmin injection during cardiovascular stress following by a second injection of an equal activity 24h later at rest. Alternatively, the two-day protocol can be conducted with the rest study as the initial study. If the stress study is performed as the first study then the image acquisition can start 15–60min after the injection. On the next day, an injection of similar dose as in the first day (500–600 MBq) is administered to the patient

![Figure T.57](image.png)
Two-day protocol

at rest with an image acquisition starting 60–90 min post-injection. The advantage of a stress/rest combination compared to rest/stress is that if the first stress study is diagnosed as ‘normal’ then the rest of the study may be omitted.

If the rest study, however, is performed as the first study, a dose of 500–600 MBq is administered and the image acquisition starts about 60–90 min after. The stress study is performed on the following day with a 15–60 min time interval between the injection of 500–600 MBq and the imaging.

From the logistic clinical viewpoint, having patients undergoing imaging on two separate days may sometimes be inconvenient and impractical, which results in increasing costs and a delay in the delivery of final information to be used in patient management. Acquiring information from both studies on a single day is therefore desirable in many cases.
Ultra-fine focus
(Diagnostic Radiology) Some special x-ray tubes are produced with very small focal spot. These tubes are either called ultra-fine focus tubes or microfocus tubes. They are mainly used in mammography, angiography or macroradiography (rarely used method these days). The ultra-fine focal spot of these tubes is normally equal or smaller than 0.1 mm effective focal spot size (also called 0.1 nominal in the IEC documents). These tubes produce sharp images with extremely high spatial resolution, but are only used for low-power exposures (otherwise the small focus will be overloaded). Some contemporary x-ray tubes use electrical system that focuses the thermal electrons, thus producing variable focal spot size, including ultra-fine sizes.

Related Articles: Anode, Cathode, Focal spot, Macroradiography

Ultrasmall particles of iron oxide (USPIO)
(Magnetic Resonance) USPIOs are a widely used type of superparamagnetic iron oxide contrast agent.

The most commonly used form of USPIO is an agent with the generic name ferrumoxtran-10, which consists of <30 nm sized particles coated with dextran. These particles are taken up by macrophages in the lymph nodes, spleen and bone marrow. Their main emerging clinical application is in differentiating cancerous from normal lymph nodes. Normal nodes take up the agent and hence appear dark on an MR image (since it is a negative contrast agent), while cancerous nodes remain bright allowing diagnosis even if they are morphologically indistinguishable from healthy nodes. There is a potential role, for example in diagnosing lymphatic metastasis in breast cancer. In Figure U.1, signal loss following administration of ferrumoxtran-10 demonstrates uptake of the agent, indicating that the lymph node is healthy.

Unlike larger superparamagnetic particles, USPIOs are generally too small to accumulate in the reticuloendothelial system, and so have a long half-life in blood. Consequently, they are promising candidates for intravascular (or blood pool) agents for use in MR angiography.

Related Articles: Negative contrast media, Superparamagnetic iron oxide, Superparamagnetic particles, Magnetic resonance angiography (MRA)

Ultrasonic field
(Ultrasound) The ultrasonic field is the spatial extent of ultrasound energy from a source. The transducer can be modelled as a summation of an infinite number of point sources, and the total field is then the summation of the field from all these sources. Rather arbitrarily, the field is usually divided into a near and a far field, separated at a distance, which corresponds to the position of the last maximum of the near field. For a circular plane transducer, this occurs at an axial distance of \( a^2/\lambda \) from the surface (where \( a \) is the radius of the transducer and \( \lambda \) is the wavelength).

Ultrasonic output
(Ultrasound) Several parameters are commonly used to describe the ultrasonic output. For safety in diagnostic ultrasound, the two indices mechanical index (MI) and thermal index (TI) are used to describe the risk with the chosen control settings. These two indices depend on the parameter's frequency and peak negative acoustic pressure (MI) and acoustic power (TI). The parameters are most often

![Figure U.1](image-url)

FIGURE U.1 Normal axillary lymph node before (a) and after (b) intravenous administration of ferrumoxtran-10 contrast agent.
adjusted automatically depending on which transducer and which type of diagnostic measurement that have been chosen. However, the operator can affect the total power and the peak negative acoustic pressure with the output power control, which varies the amplitude of the electrical transmit signal to the transducer. If the electrical transmit signal is increased, the transducer produces higher intensity sound waves, and the sensitivity of the scanner is increased.

Related Article: Acoustic power

Ultrasoundography

(Ultrasound) Ultrasoundography describes the use of ultrasound, specifically medical diagnostic ultrasound. It is often abbreviated to sonography. The term ‘ultrasound’ describes sound frequencies, which are inaudible to the human ear, that is above 20kHz. Typically, the frequencies used range from 2 to 20mHz. An advantage of ultrasound over other diagnostic imaging modalities is that it is relatively inexpensive and is safe; it does not use ionising radiation and does not usually require contrast agents. An often-quoted advantage is that it is user dependent, meaning the acquisition and interpretation of ultrasound images depend upon the training and skill of the operator.

Ultrasound

(Ultrasound) Ultrasound is sound above the threshold of human hearing. While this varies between individuals, ultrasound is generally described as sound over 20kHz, the upper limit of hearing for healthy young adults.

Ultrasound occurs in the natural world (bats use frequencies up to 100kHz) and has extensive industrial applications including cleaning, flaw detection and sonochemistry.

Diagnostic ultrasound uses frequencies from 1 to 20MHz for imaging and Doppler applications although specialist single-element transducers have been used at frequencies up to 40MHz. Biomicroscopy transducers up to 75MHz have been constructed for imaging the anterior segment of the eye.

At higher power, ultrasound is also used for medical therapy:

- Physiotherapy ultrasound devices operate at higher intensities than diagnostic systems and at frequencies from approx. 0.8 to 3 MHz.
- High-intensity focussed ultrasound (HIFU) uses focussed transducers operating from 0.5–3 MHz to destroy pathogenic tissue by heating and cavitation. Time-averaged intensities are typically 5,000–25,000W/cm² (compared with diagnostic ultrasound <1 W/cm²).

Ultrasound safety

(Ultrasound) Ultrasound is widely used in healthcare for diagnostic and therapeutic applications. The term ‘ultrasound safety’ encompasses the design and operation of ultrasound systems to ensure no or minimal adverse biological effects in clinical use. Some users and authors suggest that the term should include the scope of ultrasound practice and diagnostic competency and argue that the greatest risk from diagnostic ultrasound lies in its inappropriate use or errors of diagnosis. While this argument has merit, the term ‘ultrasound safety’ generally refers to issues of possible biological effects.

Ultrasound is a form of energy that causes thermal and mechanical effects in tissue. There is no evidence that diagnostic ultrasound has been harmful to patients to date. Therapeutic ultrasound has higher outputs, and physiotherapy systems may use vibration or temperature to achieve its effects, but there is no evidence that this causes damage to tissue. At much higher outputs, HIFU is used to destroy tissue by heating.

For diagnostic ultrasound, the study of safety is broadly divided into the following:

- Understanding of mechanisms – predominantly heating (link) and cavitation (link) and their effects on tissue
- Understanding of the likelihood of the mechanisms occurring in vivo and the relationship of output to heating and cavitation effects
- Accurate measurement of ultrasound outputs (link)
- Epidemiological studies of those undergoing ultrasound scans

Ultrasound safety has been addressed by the US Food and Drug Administration (FDA) in association with ultrasound manufacturers to give recommendations of maximum power and intensity for diagnostic applications. They have also implemented the output display standards (ODS) of T1 and MI (link) as a guide to users as to relative output and possible associated risk. National and international societies including the American Institute of Ultrasound in Medicine (AIUM) link, British Medical Ultrasound Society (BMUS), European Federation of Societies of Ultrasound in Medicine (EFSUMB) and the World Federation of Ultrasound in Medicine and Biology (WFUMB) have all issued guidelines for the duration and outputs used in diagnostic examinations.


Ultrasound-guided brachytherapy

(Ultrasound) Brachytherapy, used in the treatment of cancers, involves the placement of small radioactive sources referred to as seeds or pellets inside the body. The source may be introduced with a catheter, needle or special applicator. Ultrasound, x-ray or CT is used to position the seed accurately in the desired location. Imaging may be subsequently used to verify the position of the source. Cancers throughout the body are treated in this fashion, a typical example where ultrasound is used is in the treatment of cancer of the prostate.

Ultraviolet radiation

(Radiation Protection) Ultraviolet (UV) radiation is the part of the electromagnetic spectrum between visible light and x-rays. The UV spectrum ranges from 100 to 400nm. It is considered to be...
non-ionising radiation and is further split into different types of UV depending on the wavelength:

- UVA ranges from 400 to 315 nm
- UVB ranges from 315 to 280 nm
- UVC ranges from 280 to 100 nm

The major source of natural UV radiation is, of course, the sun. UVA comprises 95% of solar UV that reaches the surface of the Earth. UVB has a higher effective energy but only comprises 5% of solar UV reaching the Earth’s surface. The absolute amount of UVB reaching the Earth’s surface varies considerably with altitude, latitude and by the state of the ozone layer, cloud cover and pollution. UVC has the highest effective energy but does not reach the Earth’s surface and is therefore biologically irrelevant except when artificially produced. If an individual is exposed to UVC, the radiation is absorbed in the keratin in the epidermis and can cause damage to DNA.

When the solar spectrum is depicted in terms of relative ability of each UV band to cause erythema (sunburn), UVB is responsible for 93% of the erythema and UVA for 7% – that is whilst UVB comprises only 5% of the solar spectrum, it is responsible for 93% of the erythema in sunburn.

UV radiation is produced artificially and used in industrial applications and medically in a dermatology department to treat a range of skin conditions. It is of course also used in sun bed tanning units.

Related Articles: Electromagnetic radiation, Non-ionising radiation

Unblanking

(Diagnostic Radiology) Usually, the imaging chain of an x-ray video system (including image intensifier, optics, TV tube, monitor, etc.) displays some images with slight contrast degradation at the edge of the image. This effect is known as vignetting (see the eponymous article). Additionally to it, the stop of the scanning electron beam at the edge of the image creates a small white line at this place. In order to reduce these distortions of image, some manufacturers apply over the image a black circle, known as blanking circle. This way only the signal inside this circle presents an image, and the signal outside the circle is presented with 0 (i.e. total black). This way the blanking circle diameter is slightly smaller than the image intensifier field size diameter. The blanking circle moves together with the zooming (field size). Part of the blanking circle is well seen at Figures H.13 and H.14 (down, middle-right) in the article High contrast.

Although the blanking circle presents an image well perceived by the human eye, the area outside the circle (between the outer edge of the circle and the outer edge of the filed size) still receives radiation – that is small part of the patient is x-rayed but not visualised. Due to this reason, the precise adjustment of the blanking circle is very important. Removing the blanking circle (or ‘unblanking’) is necessary for adjustment and other technical purposes, as well as for assessment of this irradiated, but not imaged part of the patient.

Related Articles: Video signal, Vignetting, Image intensifier

Underexposure

(Diagnostic Radiology) Underexposure is a condition in which an x-ray image receptor has received less than the exposure required to produce an image with the desired quality. For most medical procedures using radiation there is usually an optimum amount of exposure that produces the necessary image quality. In film radiography underexposure results in reduced film density and a possible reduction in contrast. In digital radiography, underexposure of the receptor (below the optimum value) results in increased image noise.

Radiographs produced with three different exposures are compared in Figure U.3. The one on the left appears to be over underexposed. Compared to the optimally exposed one in the centre, the underexposed radiograph has much less contrast and visibility of the anatomical structures.

FIGURE U.2 Comparing an underdeveloped radiograph to one that has been correctly processed. There is a significant loss of contrast especially in the lighter areas.
Undersampling
(Magnetic Resonance) According to the Nyquist–Shannon theorem, which is a fundamental principle of information theory, the sampling frequency of a signal to be discretised has to be twice the bandwidth (the difference between maximum and minimum frequency). It is therefore important to know the maximum frequency of the signal – which is, in medical imaging, usually given by the resolution of the image to be acquired since all transforms to \( k \)-space (i.e. the Fourier domain) are performed by means of a discrete Fourier transform. If an image signal is sampled with a readout frequency of less than the required double bandwidth, the Nyquist–Shannon theorem is violated, and undersampling occurs. The result is usually a lossy reconstruction of the signal in the spatial domain. However, undersampling also allows for faster acquisition and some sort of selective pre-filtering of unwanted frequencies; this is usually referred as bandpass-sampling.

Truncation artefacts like ghosting may appear as a consequence of undersampling. These are the results of discontinuities of the spectrum acquired – if a discontinuity is encountered, for instance, by cutting-off frequencies, or by limiting the domain of the discrete Fourier transform, high-frequency artefacts show up; this is known in general as Gibbs phenomenon, which is well known in MR imaging when dealing with anatomical areas where sudden changes in contrast occur. In electronics, Gibbs phenomenon is also known as ringing. Ring or ripple-like artefacts may also be observed in MRI (Gibbs ringing).

Undertable cassette carriage
(Diagnostic Radiology) X-ray radiography systems using x-ray tube mounted above the patient table are equipped with an undertable cassette carriage. This carriage is mounted under the anti-scatter grid of the Bucky device. The carriage has a drawer-holder that moves in/out to load/remove the x-ray film cassette (Figure U.4).

Related Articles: Radiography, Bucky table

Undertable fluoroscopy
(Diagnostic Radiology) Undertable fluoroscopy x-ray systems use x-ray tube mounted under the patient table and image intensifier (II) above the patient table (Figure U.5). These are the most often used fluoroscopic systems, as the operation of the fluoroscopy is conveniently placed over the II, what allows the radiologist/radiographer to easily guide the investigation. Another advantage of these systems is the fact that the II absorbs most of the scatter radiation from the patient. This way, the undertable fluoroscopy x-ray systems expose the staff to much lower dose from scattered radiation.
For example, in such a system, the scattered radiation dose rate is of the order of 1 μGy/min at 1 m from the patient. In comparison, an overtable fluoroscopic system will have at the same place approx. 20 times higher dose rate.

In the case of overtable fluoroscopy, the x-ray tube is in front of the patient, and it is possible to have mounted in front of it a remote-controlled pressing device (also called palpator). This devise is useful for some investigations, for example to move the barium meal in the stomach. Overtable fluoroscopy also allows for larger focus-patient distance.

The multifunctional C-arm-mounted fluoroscopic systems move the x-ray tube at various positions around the patient.

**Related Articles:** Radiography, Fluoroscopy

**Further Reading:** Coulam, C., J. Erickson, D. Rollo and A. James eds. 1981. The Physical Basis of Medical Imaging, Appleton-Century-Crofts, New York.

### Undertable radiography

_(Diagnostic Radiology)_ Undertable (undercouch) radiography uses x-ray tube mounted under the patient table. This installation is usually associated with fluoroscopic units with overtable image intensifier. These systems produce considerable less scattered radiation (from the patient) in comparison with the systems where the image intensifier is below the patient table (see the article on Undertable fluoroscopy).

**Related Articles:** Radiography, Fluoroscopy, Undertable fluoroscopy, Undertable radiography

### Unenhanced image

_(Magnetic Resonance)_ This term relates simply to an image that has not been enhanced by the use of an exogenous contrast agent. In clinical practice, it is common to collect an unenhanced ‘baseline’ image prior to administration of contrast agent. The flexibility of MRI in terms of the information content of images and the ability to manipulate contrast through pulse sequence design also means that there is a greater role for unenhanced imaging than is the case with x-rays.

**Related Article:** Contrast agent

### Ungrounded

_(General)_ Systems or devices that have no direct electrical connection to Earth.

**Related Article:** Earthing

### Uniformity

_(Diagnostic Radiology)_ In a CT image, the term ‘uniformity’ refers to the scan plane variation of CT numbers in a homogenous object, usually a water phantom. There will be some inherent variation due to stochastic noise, but the mean CT number value in regions of interest (ROIs), positioned at any point within the FOV, should be relatively constant.

Uniformity can either be assessed by plotting a CT number profile across the phantom diameter (Figure U.6) or by placing ROI at the centre and peripheral regions and comparing the mean CT number in each ROI.

The Institute of Physics and Engineering in Medicine (IPEM) recommends that the variation in mean CT number between a ROI at the centre, and one at the periphery, of a uniform phantom, should not be more than ±10 HU and ±20 HU in head-sized and body-sized phantoms, respectively.

Poor uniformity is usually caused by artefacts such as beam hardening or ring artefacts.

**Related Articles:** Beam hardening, Ring artefact

**Figure U.6** Image of homogenous water phantom showing good uniformity. (Graphs courtesy of ImPACT, UK, www.impactscan.org)

**Further Reading:** IPEM (Institute of Physics and Engineering in Medicine). 2005. Recommended standards for the routine performance testing of diagnostic x-ray imaging systems, IPEM Report 91, York, UK.

### Uniformity

_(Magnetic Resonance)_ Uniformity can be used in a number of ways in magnetic resonance imaging. The first is the uniformity of the main magnetic field and the RF field. The uniformity of the main magnetic field is an important criterion of the quality of the magnet as non-uniformities will often lead to image artefacts. The main feature that affects RF uniformity is the RF coil. Structures close to the surface can be imaged with the use of a surface coil, which is not uniform, while internal structures often use a volume coil, which provides the most uniform RF field.

Signal and SNR uniformity measures are used in quality control tests to monitor scanner performance. Uniformity of image signal looks at the variation of image signal intensity over an image. Uniformity of SNR looks at the variation of SNR over an image. These tests ensure there is a constant signal and SNR across an image slice and that there are no image artefacts present.

**Abbreviations:** RF = Radiofrequency and SNR = Signal to noise ratio.

**Related Articles:** RF uniformity, Field uniformity

### Uniformity

_(Nuclear Medicine)_ Uniformity is an imaging detectors capability of depicting a homogenous distributed flood source. There are two primary causes for non-uniformity in a scintillation camera. The first one is non-uniform detection efficiency, which arises from (1) the small difference in PM tube pulse height spectrum (or PM tube response) and (2) the position dependence of scintillation light collection efficiency, that is events occurring perpendicular to one of the gaps between two PM tubes will register a lower pulse than events originating perpendicular to the centre of a PM tube. The PM tube response can be tuned so the effect is minimised. The position-dependent efficiency is hard to compensate for. The best way is to minimise the gap between the tubes and not to use the FOV close to the physical PM tube edge.

The second cause for non-uniformity is image non-linearities, that is when straight line objects appear to be curved in the image. They occur when the X- and Y-position does not change linearly.

**Related Articles:** Beam hardening, Ring artefact

**Related Articles:** Field uniformity, Radiofrequency
with the position across the face of the detector. Nonlinearities can occur for a number of reasons, for example difference in sensitivity among the PM tubes, non-uniformity in optical light guides and PM tube or electric malfunctions.

Another case of non-uniformity is a bright ring around the edge of the image. This characteristic artefact is called edge packing and results from a small increase in light collection efficiency for events originating close to the crystal edges. Light photons can be reflected at the crystal edge surface and then registered in one of the PM tubes. Regions demonstrating this effect are often masked because they occur in the outer region of the FOV leaving only the useful FOV.

In planar emission imaging, the effect of detector non-uniformity is relatively low (compared to tomographic emission imaging). Acceptable variation in count rate in a flood-field image can be ±10% or more. In tomographic imaging, non-uniformity can lead to characteristic ring-artefacts. It is therefore customary to use a uniformity correction on patient data. Uniformity should be measured as a part of the routine quality control. Data from the quality control are then used to update the uniformity correction. An effective uniformity correction can justify the use of smaller PM tubes and thinner light guides to improve the event localisation and intrinsic spatial resolution.

**Related Article:** Intrinsic flood field uniformity


**Unipolar output**

*(General)* A unipolar output normally refers to an electronic device capable of accepting or providing an output, which has one polarity (0 to +X volts or 0 to –X volts). This is a common form of output and is easy to understand and process when the parameter it is representing is also unipolar (Figure U.7).

When analogue-to-digital converters (ADC) are used, unipolar ADCs accept only unipolar signals and output only an unsigned binary number from zero to some predefined maximum.

When a parameter to be represented electronically varies through both positive and negative values, a bipolar input and output may be necessary, though this sometimes poses design problems.

If a unipolar device (such as an amplifier or ADC) is to be used with a bipolar input signal, a reference voltage must be added to ‘offset’ the signal prior to digitisation. A common standard is to add a reference voltage equal to half the maximum input range of the ADC, thereby providing an output in ‘offset binary’.

In digital electronics, the term ‘unipolar output’ may have a different meaning, referring to the inability of the logic device to both actively pull up and pull down any attached load, which may be connected. In such cases, it is important to make sure some other method such as ‘pull up’ or ‘pull down’ resistor is attached to that output to ensure that both logic states are detectable at the output.

**United Nations Scientific Committee on the Effects of Atomic Radiation (UNSCEAR)**

*(General)* The UNSCEAR is part of the United Nations (UN) and reports directly to the General Assembly.

UNSCEAR was founded in 1955 with the mandate to assess and report levels and effects of exposure to ionising radiation, within the UN system. The UNSCEAR reports are considered as basis for the evaluation of radiation risk and therefore for the implementation of adequate radiation protection.

The UN General Assembly designates the countries from where the scientists, who are going to be members of the committee, are coming from. The work program is also approved by the UN General Assembly and typically extends over a period of 4–5 years.

The secretariat is based in Vienna and is functionally linked to the United Nation Environmental Program (UNDP). The main function of the secretariat is to organise the annual sessions and to prepare the documents for the committee scrutiny. The secretariat collects relevant data submitted by UN Member States, international and non-governmental organisations and engages specialists to analyse data, study the scientific literature and produce scientific reviews. The scientific reviews are then submitted to each UNSCEAR session, and at the end of the process, the substantive reviews are published.

The committee has published important scientific documents, starting with the two reports produced in 1958 and 1962, which were the basis for the signature of the Partial Test Ban Treaty on the prohibition of nuclear weapon tests in the atmosphere. Other important reports were published on the assessment of radiation exposures and health effects following the Chernobyl accident in 1986. In fact, a first report was published in 1988 on acute radiation effects in emergency workers and on the global exposures. A more detailed assessment of radiation levels and effects was published in 2000.

The latest UNSCEAR program includes the evaluation of risk from exposure to radon, epidemiological studies on radiation-induced cancer and non-cancer effects, radiation effects on the immune system, etc. The latest report will be published next year.

UNSCEAR is a member of the Inter-Agency Committee on Radiation Safety (IACRS).

**Hyperlink:** [http://www.UNSCEAR.org](http://www.UNSCEAR.org)

**Universal wedge**

*(Radiotherapy)* A universal wedge is one of a given angle that is fixed in a beam for all beam widths up to a given size. See Figure U.8.

When a small field is used, only a small thickness of wedge is needed to produce the tilt in the isodose plot. The remainder of the wedge thickness attenuates the beam and results in a reduction of dose rate.

**Unsealed source**

*(Nuclear Medicine)* Unsealed source, or open source, refers to a radioactive source, which is not encapsulated or otherwise contained. From a radioprotection point of view, an unsealed source can lead to contamination if not handled properly. In nuclear medicine, radionuclide solutions and radiocompounds are unsealed sources.

Radiotherapy when using an unsealed source is also referred to as unsealed source therapy.

The opposite of an unsealed source is a sealed or closed source.

**Related Article:** Sealed source
Unsharp masking

Unsharp masking is a software method used in digital subtraction angiography (DSA). The method is specially useful for cardiac angiographic examinations. In this method, the mask image (the one to be subtracted from the image with maximal contrast) is formed as a superimposed image. As this mask is superimposed, it is with unsharp contours (over-smoothed image). On the contrary, the image with maximal contrast represents the cardiac muscle at specific moments of the cardiac cycle – that is the contours of each image are different. When the unsharp mask is subtracted from these images with varying contours, the resultant subtracted images have similar blurring of their contours/edges, what facilitates the visual representation of the movements in the subtracted image. As a whole, the unsharp masking results in an overall image noise reduction. Different manufacturers apply different image processing algorithms for unsharp masking.

An example is shown in the following. Figure U.9 shows a DSA image of the heart (subtraction between the images from exposure 40 and exposure 25). With this fixed subtraction, two images of one and the same vessel are presented (light and dark). Figure U.10 (from the same DSA examination) shows unsharp mask subtraction between image 40 and an integrated sum of several other images. With this unsharp masking, the image of the vessels is clearly shown as one structure.

Unsharpness

Unsharpness is a characteristic of an image resulting from blurring. In such an image, structures, objects and edges appear to be ‘unsharp’.

Related Articles: Detail resolution, Geometric unsharpness

Use factor

Radiotherapy facilities must be properly shielded in order to limit radiation exposures to members of the public and employees to an acceptable level. The NCRP Report No. 151 provides technical data and recommendations regarding the design and installation of structural shielding for high-energy x-ray and gamma-ray radiotherapy facilities. According to this report, for calculation of the design of primary and secondary barriers, it is necessary to consider various factors such as workload W, use factor U and occupancy factors T. The barrier thickness (B) can be calculated as follows:

\[ B = \frac{Pd^2}{WUT} \]

\[ n = -\log(B) \]

\[ t = TVL_1 + (n-1)TVL_{eq} \]

- \( P \) – shielding design goal (expressed as dose equivalent) beyond the barrier
- \( d \) – distance from the x-ray target to the point protected (m)
- \( U \) – use factor
- \( T \) – occupancy factor for the protected location of fraction of the worktime that a person is present beyond the barrier
- \( W \) – workload or photon-absorbed dose delivered at 1 m from the x-ray target
- \( t \) – the barrier thickness
- \( TVL_1 \) – first tenth-value-layer
- \( TVL_{eq} \) – equilibrium tenth-value-layer

The use factor U or beam direction factor is the fraction of time during which the radiation under consideration is directed at a particular barrier. The use factor is always equal to one for scattered and leakage radiation because these radiations impinge on the barrier for all orientations of the primary beam.

FIGURE U.8 A universal wedge where the wedge is fixed in the centre of the beam and the field can be opened to just about any size. The unused segment only serves to reduce the beam dose rate.

FIGURE U.9 DSA image of the heart vessels without unsharp masking. (Graphs courtesy of EMERALD project, www.emerald2.eu)

FIGURE U.10 DSA image of the heart vessels with unsharp masking. (Graphs courtesy of EMERALD project, www.emerald2.eu)
Typical use factors are

<table>
<thead>
<tr>
<th></th>
<th>( U = )</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Full use</td>
<td>( U = 1 )</td>
<td>Floors of radiation rooms except dental installations, doors, walls and ceilings of radiation rooms exposed routinely to the primary beam</td>
</tr>
<tr>
<td>Partial use</td>
<td>( U = 1/14 )</td>
<td>Doors and walls of radiation rooms exposed routinely to the primary beam, also floors of dental installations</td>
</tr>
<tr>
<td>Occasional use</td>
<td>( U = 1/16 )</td>
<td>Ceilings of radiation rooms not exposed routinely to the primary beam. Because of the low use factor, shielding requirements for a ceiling are usually determined by secondary rather than primary beam considerations</td>
</tr>
</tbody>
</table>

**Related Articles:** Tenth value layer (TVL), Occupancy factor, Treatment room, Maze

**Further Reading:** NCRP (National Council on Radiation Protection and Measurements). 2005. Structural shielding design and evaluation from megavoltage x- and gamma-ray radiotherapy facilities, NCRP Report 151, Bethesda, MD.

**Useful beam**

(Radiotherapy) The x-rays produced when the electron beam is incident on the target are predominantly in the forward direction (towards the patient). However, while these x-rays are of benefit in forming a beam (following some further modifications and shaping), which may be of use for patient treatment, the x-rays produced in the other directions are of no benefit. Therefore, a significant amount of shielding is in place in the head of a linac to reduce the dose from these other x-rays and provide a clearly defined treatment field. This treatment field is known as the useful beam, that is the beam that is used to irradiate the patient.

**Related Articles:** Collimation, Collimator, Primary collimator

**Useful field of view (UFOV)**

(Nuclear Medicine) The FOV is the maximum image size that the image system is capable of imaging. Only a portion of the FOV is suited for imaging or activity quantification. This portion is referred to as the useful field of view. The UFOV is determined by irradiating a detector with a uniform flood source. The UFOV is the area where the uniformity does not deviate more than 10% from a ROI placed in the central area of the image field. The UFOV is typically 80% of the total FOV. Using a scintillation camera, artefacts close to the edges prevent any use of that information. In PET, the UFOV is determined by the number of opposite elements used in the fan-beam acquisition.

During the quality control of the camera system, the resolution, linearity and uniformity are measured over the UFOV and/or the central FOV.


**Related Articles:** Centre of field of view (CFOV), Field of view, PET, SPECT

**User interface**

(Diagnostic Radiology) The user interface is a very broad term associated with any human information exchange with a machine. In computers, the user interface is the way graphical, or other information is delivered by the software to the operator and, respectively, the way the operator submits his commands to the software. Often the user interface is called also control panel (physical or software).

**USPIO**

(Magnetic Resonance) See Ultrasmall particles of iron oxide (USPIO)

**UV radiation**

(Radiation Protection) See Ultraviolet radiation
**Vacuum**  
*(Diagnostic Radiology)* A vacuum is an empty space. The name is derived from the Latin, *vacuus*, meaning empty. Most electronic tubes, including x-ray and image intensifier tubes, are evacuated, so there is no air or other gases to impede the passage of electrons. Vacuum is measured as pressure in pascals (SI unit, 1 Pa = 1 N/m²) or in bars (1 bar = 100 kPa). For example, the vacuum in an x-ray tube is minimum 10⁻⁶ mbar. It is necessary so as not to impede the flow of electrons, usually from a cathode to an anode.

There is never an absolute vacuum, and the quality of a specific vacuum is defined by the pressure of gases, such as air, that are in the space.

**Valence band**  
*(Nuclear Medicine)* In semiconductors and isolators, the highest energy band with electrons present at absolute zero is called the valence band. The bands are separated by regions with forbidden energy states, which electrons cannot occupy. The band following the valence band (i.e. a band occupied by electrons with higher energy) is called the conduction band. As the name implies, electrons in the conduction band take part in the conduction of electricity. Electrons can be excited up to the conduction band via thermal excitation. In metals, these two bands are not separated, and metals are therefore generally good conductors.

*Related Article:* Conduction band

**Valve tube rectifier**  
*(Diagnostic Radiology)* Valve tube rectifier, also known as vacuum rectifier, is a rectification circuit that uses vacuum tube diodes. Such rectifiers were often used in high-voltage generators of x-ray equipment, but nowadays are replaced by power semiconductor diodes (Figures V.1 and V.2).

*Related Article:* Rectifier

**Van Cittert–Zernike theorem**  
*(Ultrasound)* As first applied in optics, the Van Cittert–Zernike theorem describes the spatial covariance of the field produced by an incoherent source. It states that the spatial covariance of the field, sensed at two points $X_1$ and $X_2$ of an observation plane, is equal to the Fourier transform of the source aperture function.

In ultrasound imaging, we may consider the sound scattered back from a multitude of scattering objects as the incoherent source and our observation plane, the transducer face. The energy distribution radiated in the focal plane from a rectangular aperture is a sinc-function squared that is the Fourier transform of the aperture function. Thus, sound scattered back from objects in the focal plane will have an energy distribution in this form, so the spatial coherence at the aperture is, according to the theorem, the Fourier transform of this sinc²-distribution. This, in turn, will be a triangular function with a base being twice the size of the aperture. This is equal to say that the spatial covariance of the backscattered field is proportional to the autocorrelation of the transmitter aperture function. A consequence is thus that the longer the aperture, the wider the spatial covariance of speckle in the focal plane. However, away from the focal zone, the decorrelation is more rapid.

This theorem is useful for locating regions where the speckle is well correlated for speckle-tracking applications. It can also be used in aberration correction as the coherence of the backscattered field could be used to obtain information about the illuminating beam width.


**Variable thickness transducer**  
*(Ultrasound)* See *Slice thickness*
Variable transformer

(General) A variable transformer is a form of transformer where the number of windings associated with the output circuit can be varied with respect to the number associated with the input.

Such transformers are usually autotransformers, having only one common winding for both input and output, but with the output connected via one of a selectable set of ‘taps’ or connections (Figure V.3).

In this manner, an efficient device can be made that allows the selection of a wide range of AC output voltages for a given fixed AC input voltage.

Most variable transformers are used in mains power circuits, and it should be noted that being autotransformers, they provide no electrical isolation between input and output.

Related Article: Transformer

Variable-aperture beam restrictor

(Diagnostic Radiology) See Beam restrictor

Vector array

(Ultrasound) A vector array is another term for phased array transducer, whereby scan lines are fired from a linear array in different directions, allowing a sector to be imaged (Figure V.4). In some phased/vector arrays, the image at the transducer is depicted as a point, in others a short straight line and in others a short tightly curved line.

Related Article: Phased array transducer

Veiling glare and contrast

(Diagnostic Radiology) Veiling glare is observed in optical systems. It is caused by stray light reaching the sensor of an imaging system, causing distortions in imaging performance. The main source of veiling glare is reflections between the surfaces of the lens caused by reflections of the object itself or other out-of-field objects. The resulting image is with decreased contrast in certain areas affected by the veiling glare.

Veiling glare in x-ray fluoroscopy is seen at the output window of the image intensifier (II). Part of the light emitted by the output phosphor is reflected by the glass of the output window. This light bouncing reduces the overall contrast of the image. Other scattering effects in the input phosphor and the accelerated electron beam could also lead to reduced contrast due to veiling glare.

If a radio-opaque material (lead disk) is placed in front of the II ideally, the light intensity of its image should be zero; however, due to veiling glare, this intensity has some value. Contrast reduction due to veiling glare can be measured by placing a lead disk at the centre of the II (usually the diameter of the disk is 10% of the diameter of the II). The light intensity at the centre of the II (behind the disk) is measured and is compared with the same light intensity without such lead disk.

The ratio (light intensity without disk)/(light intensity with disk) gives indication about the veiling glare. The result of this measurement is called contrast ratio and normally a good II should have contrast ratio >30.

Related Article: Image intensifier

Velocity encoding (VENC)

(Magnetic Resonance) In velocity encoding (also known as flow encoding), dedicated gradient waveforms are applied, which result in a velocity-induced phase shift \( \Delta \Phi \). The design of the gradient waveforms determines the maximal velocity that can be encoded (the so-called VENC), which is defined as the velocity at which \( \Delta \Phi = 180^\circ \). Hence, \( v = \text{VENC} \cdot \Phi_{\text{net}}/180 \).
In standard velocity mapping sequences, the VENC needs to be defined prior to sequence start, and therefore a general knowledge of expected maximum velocities in the studied vessel is necessary to optimise contrast-to-noise (CNR) in the velocity maps. Selection of a too low VENC (a VENC lower than the actual maximum velocity encountered) results in phase wraps in the velocity map, since the velocity-induced phase angle is only unambiguously defined in the interval $[-180^\circ, 180^\circ]$.

Conventionally, the VENC value (often in units of cm/s) is part of the sequence protocol at the scanner and can be chosen within certain limits. A decrease in VENC (increase in velocity sensitivity) requires higher velocity-encoding gradients for a fixed TE and therefore the lower the VENC.

**Related Article:** Velocity mapping

**Velocity mapping**

*(Magnetic Resonance)* Velocity mapping, also known as phase-contrast MRI (PC-MRI), is a MR technique for the measurement of the velocity (in one or more directions) of liquid or tissue motion.

When a spin moves with a constant velocity within a gradient field, it will acquire a net phase offset equal to

$$\phi_{\text{moving}} = \int_0^t \omega(t) \gamma G(x, t) v \, dt$$

where

- $\omega$ is the Larmor frequency of the spin
- $\gamma$ is the gyromagnetic ratio

Similarly, a non-moving spin will acquire a net phase offset of

$$\phi_{\text{static}} = \int_0^t G(x, t) \gamma x_0 \, dt$$

If a subsequent gradient waveform of opposite polarity is applied immediately after the first gradient, the net phase offset of the static spins will be zero, but the moving spin will end up with a net phase offset proportional to its velocity. Such a gradient waveform is called a bipolar gradient and is often used in velocity mapping (Figure V.5).

Phase offset of spins in velocity mapping can be caused by other factors than the velocity of the spins, for example field inhomogeneities and Eddy currents. These effects can be compensated by repeating the experiment with a bipolar gradient of opposite polarity or with another gradient strength of the bipolar gradient. The subtracted image will ideally only reflect a velocity-induced phase shift.


**VENC (velocity encoding)** *(Magnetic Resonance)* See Velocity encoding (VENC)

**Venetian blind artefact** *(Magnetic Resonance)* Venetian blind artefacts are artefacts affecting magnetic resonance angiography. MRA visualises flow in blood vessels with the blood flow imaged in sections called slabs. Slowly flowing blood can become saturated and produce signal drop-out, which increases with slab volume. One method of improving problems found with this technique is to use multiple overlapping thin slab acquisition (MOTSA). This involves reducing the saturation effect seen in MRA by decreasing the thickness of the slabs imaged, but maintaining the overall volume by increasing the number of slabs.

The venetian artefact occurs at the boundaries of these slabs, where they slightly overlap. The background tends to be brighter at the top of the slab compared with the bottom due to an increased flip angle. A variation in signal response across the slab boundaries can then be seen in the phase-encode direction.

**Abbreviations:** MOTSA = Multiple overlapping thin slab acquisition and MRA = Magnetic resonance angiography

**Video camera tube** *(Diagnostic Radiology)* The video camera tube (VCT; also called television pickup tube or TV camera tube) is an important element of the x-ray television (x-ray fluoroscopy). The VCT is a cylindrical vacuum device with base approx. 3 cm (∼1 in. diameter) and length of approx. 15 cm.

The front of the VCT is attached to the output optics of the II and takes directly the image from it. Figure V.6 shows a block diagram of VCT. The VCT front consists of glass entrance window, transparent conductive layer (signal plate) and photoconducting layer (target plate). When light from the II passes through the glass and the transparent signal plate, it reaches the target plate.

![Figure V.5](image-url)

**Figure V.5** (a) A bipolar gradient with amplitude $G$. In practice, gradients have a trapezoid shape due to gradient rise times. The gradient lobes can also be separated in time. (b) Corresponding phase shift of a moving and a stationary spin.
The material of the target plate is the most important element of the VCT and often the whole tube is named according to it (e.g. if the target is lead oxide, PbO, the camera type is plumbicon). The vidicon camera uses target of antimony sulphide (Sb$_2$S$_3$), which is suspended as microglobules over a matrix made of mica. This way, each microglobule (approx. 0.001 in. in diameter) is electrically isolated from the other. The target is an isolator if not lighted. When it is lighted, it reduces proportionally its electrical resistance.

The cathode of the VCT is often a heated filament producing thermal electrons (electron gun). These electrons are accelerated through electrodes (approx. 250 V) towards the anode, which is in front of the target plate. A steering system of deflection coils moves the electron beam to scan the surface of the target plate. Different amount of electrons passes through different microglobules of the target plate, depending on their resistance – that is light level over these microglobules. This way, different current passes through the respective regions of the positive signal plate (which is a conductor). This current is proportional to the light levels over the target and forms the video signal from each element of the surface of the target plate. The VCT is an analogue device, but the microglobules could be regarded as pixels (what would be the case with a CCD camera). The amplified video signal is used to produce an analogue image over a TV monitor.

The VCT and its associated optics and electronics form the TV camera assembly (Figure V.7).

**Related Articles:** Video signal, Plumbicon, Vidicon

**Further Readings:**

**Video detector**
*(Nuclear Medicine)* Charge-coupled devices (CCDs) have high quantum efficiency for optical photons and are thus very well suited for detecting light from scintillators. When cooled, they have almost no dark current. Also, noise in the readout electronics can be reduced by built-in electron magnification or parallel readout.

Using CCDS in small animal imaging gives the possibility of building small scintillation cameras based on optical lens systems projecting the scintillation light onto the CCD. Such systems can also be used for optical imaging of bioluminescence or fluorescence.

**Abbreviation:** CCD = Charge-coupled device

**Related Articles:** CCD, Scintillation camera, Fluorescence, Bioluminescence

**Video recorder**
*(Diagnostic Radiology)* Video recorders were used in the past to record analogue fluoroscopic information. The video signal from the x-ray TV camera was recorded either on tape (videotape recorder) or on hard disk and from it to CD or DVD (video recorder). These systems were used as replacement of cine film, as video recording is made with significantly lower radiation dose (direct recording of fluoroscopy requires approx. 10 less dose than cinefluoroscopy). However, the resolution of the video recording is significantly low. Contemporary digital fluoroscopic x-ray systems record directly on hard disk, the digitised signal from the detector (TV camera with analogue to digital converter, CCD camera, flat panel detector, etc.). The fluoroscopic file is then recorded on CD/DVD or other media to allow visualising at another place.

**Related Articles:** Fluoroscopy, Digital fluoroscopy, Cinefluoroscopy


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**FIGURE V.6** Block diagram of video camera tube.

**FIGURE V.7** TV camera with removed video camera tube (vidicon).
**Video signal**

*(Diagnostic Radiology)* The video signal from the TV camera of an II chain represents important information for the II function and the radiation dose during fluoroscopy. The video signal is directly related to the II input dose rate (air kerma rate) and to the image contrast.

The video signal is measured with a calibrated oscilloscope, usually connected directly to the output of the TV camera tube (usually with 75 ohms terminating resistor). One video signal represents one horizontal line from the TV raster. The signal is called composite video signal as it includes elements related to the signal and to the synchronisation of the monitor.

Figure V.8 shows a typical composite video signal. It begins with a synchronisation pulse (marking the beginning of the horizontal raster line). The difference between the black level (dark screen) and the white level (full brightness) is related to image contrast. The amplitude of the white is related to the II input dose rate (at the end of the white signal is the effect of vignetting associated with signal distortion). The blanking level is below the black level as it represents the completely black circle around the image. The uneven parts of both black and white levels represent, respectively, the black noise and the white noise.

The image on the analogue TV monitor screen, carried by the video signal, consists of a number of such composite video signals (one for each horizontal line). The overall image of a normal resolution TV monitor has a number of horizontal lines (raster lines) depending on the standard (e.g. 625 lines in the European Union and 525 lines in the United States). Each horizontal video signal ends with a horizontal sync pulse as on Figure V.8, which moves the beam one step down and returns it at the beginning of the next horizontal line. At the end of the last horizontal signal (down right corner of the screen), a vertical sync pulse moves the beam back at the upper left corner of the screen to start a new image. Not all horizontal lines carry information. About 10% of those (mainly the most upper and most lower ones) are not used – for example in the United States, standard approx. 480 lines (from 525) are forming the image on the monitor. Increasing the number of raster lines on the monitor leads to better image resolution (e.g. using 1125 lines instead of 525). In digital LCD monitors, this resolution depends on the pixel matrix.

**Related Articles:** Image intensifier, Video camera tube, Unblanking

**Vidicon tube**

*(Diagnostic Radiology)* The vidicon TV camera tube (also called hivicon) has been developed by the RCA. This camera has a target with photoconductive layer of antimony sulphide (Sb$_2$S$_3$), suspended as globules on a mica matrix. Each globule is insulated from its neighbours. These globules behave as tiny capacitors. This light-sensitive layer (usually with 1 in. diameter) is sometimes called retina. The operation of the camera is explained in the article Video camera tube. According to the light intensity at the target of the camera, an electrical charge pattern is formed over the photoconductive layer. When the target is scanned by an electron beam, the variously charged micro-areas are discharged, respectively, and the varying discharging current (proportional to the charge of the layer, hence to the intensity of the incoming light) forms the video signal.

The change of the discharge current is related to change of the resistance of the micro-areas on the target (when Sb$_2$S$_3$ is illuminated, it conducts electrons; when dark, it behaves as an insulator). This means that the amplitude of the signal is proportional to the intensity of the light.

The sensitivity of the vidicon can be changed by changing the voltage applied to the target (in comparison, the plumbicon TV camera has fixed sensitivity). This way, TV systems using vidicon may have a circuit that monitors the light intensity level and, on this basis, operates automatic gain control. This means that this TV tube can operate at various conditions of light. Also, this means that vidicon tubes can be used in system without automatic brightness control.

Vidicon TV tubes have high dark current and are also more inert than plumbicon tubes. That is, they keep the old image for some time (what presents an effect of residual image, known as ghosting). This slow response is with time constant of the order of 1/5–1/10 s. This has two effects. From one side, the camera cannot be used to image rapid movement (i.e. heart movement), as it will blur the image. From other point of view, this inertia creates an averaging effect, which minimises the noise in the image.

The vidicon camera has a characteristic curve with gamma 0.7. This means that the camera will have a wider contrast range (latitude), but the picture will be with lower contrast (contrast loss) than TV tubes with gamma 1 (as is plumbicon). However, vidicon is less expensive than plumbicon.

Vidicon is better suited for high-contrast fluoroscopic examination of less dynamic objects (e.g. barium examination of stomach).

CCD-type cameras and flat panel detectors gradually replace the fluoroscopic x-ray systems with TV camera tubes.

**Related Articles:** Video camera tube, Superorthicon, Plumbicon


**Hyperlink:** The Cathode Ray Tube site http://members.chello.nl/~h.dijkstra19/page4.html

**Viewing angle**

*(Diagnostic Radiology)* The viewing angle of an image display describes the ability of the device to be viewed at various angles by
Contrast ratio is defined as the ratio of the brightness seen through the LC cell at different viewing angles as the light passes through a greater length of liquid crystal at increased angles, and thus, the phase difference of the two light components will differ leading to a difference interference at the polariser, see Figure V.10.

To reduce the effect that birefringence has on viewing angle, modern TN LCDs include a retardation film or discotic film. This is placed between the LC cell and the analysing polariser, which increases the viewing by rotating the angle of polarisation of light entering the polariser compensating for the retardance induced by the LCD through having the opposite birefringence.

The use of the first type of liquid crystal displays was restricted by low contrast ratio, narrow viewing angle and slow response time. To overcome these issues, different manufacturers have developed LCDs based upon several different molecular alignment and electrode patterns. These are TN, in-plane switching (IPS) and vertically aligned (VA) LCD.

Related Articles: Liquid crystal display (LCD), Active matrix flat panel liquid crystal display

Vignetting (Diagnostic Radiology) Vignetting is an optical term that refers to the gradual loss of brightness and/or saturation in the periphery of an image, compared to its centre. The resulting image is with dark corners, often showing the sides of the lens. Although considered as a special effect in photography, vignetting has a negative influence over the diagnostic image. Vignetting source could be mechanical, optical, natural or pixel.

Vignetting in x-ray fluoroscopy is seen as brightness reduction at the periphery of the image. Its primary causes are the reduced precision of the lens-like effect of the accelerating electro-magnetic filed inside the II; the reduced light production at the periphery of the slightly curved input screen of the II; the reflection of light around the output phosphor. Additionally, there may be vignetting due to the optical tandem optics between the II and the TV tube (if such optics is used). The summary effect of these is usually not more than 25% brightness reduction. Vignetting is well seen at the graph of a video signal (in the eponymous article; Figure V.11).

Related Articles: Video signal, Unblanking

Virtual source position (Radiotherapy) Electron beams appear to originate from a point in space that does not coincide with the scattering foil or the accelerator exit window. The term virtual source position was introduced to indicate the virtual location of the electron source.

Virtual source point or position is relevant to electron beams since they have passed through a scattering filter. This will change the beam from its well-defined collimated shape to one which diverges. The degree of divergence will be energy dependent and will mean that electron beams with different energies will appear to have originated at different source positions.

Source position is an important factor when calculating the change in output factor for extended FSD treatments.
Virtual source position is also referred to as apparent or effective source position or point.

**Abbreviation:** FSD = Focal to surface distance

**Related Articles:** Apparent source position, Effective source point

**Viscosity**
*(Nuclear Medicine)* Viscosity is a measure of the resistance of a fluid when subjected to external stress, that is being deformed. The higher the resistance to flow, the higher the viscosity. Oil, for example has a higher viscosity than water.

**Visual grading analysis (VGA)**
*(Diagnostic Radiology)* VGA is a method for evaluating clinical image quality by assessing visibility of important anatomical or pathological structures. This method is characterised by simplicity but enabling to quantify subjective opinions and to make them amenable to analysis.

VGA can be performed in two major ways: with relative or absolute grading. In relative VGA, the quality of an image or a particular part of the image is compared with a reference image, and a grading is given depending on whether the quality of the image is better or worse than the reference image. In absolute VGA, the scores are given on an absolute scale (i.e. fulfilment of image criteria, e.g. European quality criteria), typically consisting of four to five scale steps ranging from ‘very bad’ to ‘very good’.

In relative VGA, the observer compares an image with a reference image and gives a statement of the relative visibility of the structure. The scale typically consists of five steps ranging from ‘much worse’ to ‘much better’. Prior to analysis, the scale steps in a VGA study are most often converted to numerical values. For example, in an absolute VGA study with four scale steps, to the lowest scale step may be attributed the number 1 and to the highest scale step the number 4. In a relative VGA study with five scale steps, to the lowest scale step may be attributed the number −2 and to the highest +2.

The results of a VGA study can be summarised in the VGA score (VGAS) as follows:

$$\text{VGAS} = \sum_{i}^{n_s} \frac{S_i}{N_iN_o}$$

where

- $S_i$ are the given individual scores for observer $O$ and image $I$
- $N_i$ is the total number of images
- $N_o$ is the total number of observers.
VGA with reference images is very suitable for image quality studies with modern workstations, where two, three and sometimes even more monitors are used.

**Related Articles:** Anatomical noise, Visual grading characteristics (VGC)


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**Visual grading characteristics (VGC)**

(*)Diagnostic Radiology*) VGC analysis is a method for evaluating clinical image quality, based on visual grading (e.g. image criteria [IC] study or absolute VGA) data and similar concepts as developed for ROC analysis. VGC can be interpreted as a repeated image criteria scoring, where the observer changes his threshold for the fulfillment of each criteria in a similar way as the scale steps in an ROC study are used by the observer to state the confidence of each positive/negative decision. So the probability distribution of the images from each modality is sampled.

VGC is conceptually different from VGA – VGC corresponds to the observer grading his confidence in the fulfillment of an image quality criterion whereas VGA corresponds to the observer grading his opinion about the visibility of a certain structure.

VGC analysis can be used directly on the image quality criteria defined by the European Commission or on other radiographic quality criteria – giving statements of the required levels of reproduction for certain anatomical landmarks – without the need for extracting the relevant structures from the criteria and grading the visibility of these structures. Furthermore, the grading task in VGC is not limited to normal anatomy – grading of image criteria based on pathology may also be used.

**Related Articles:** Receiver-operating characteristics (ROC), Anatomical noise


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**Voltage**

(*)General*) The voltage refers to the electrical potential of a point in a circuit with respect to some other reference level. Unless specified, the reference used is commonly taken to refer to ‘ground’, the earth potential provided by the mains supply or the chassis potential of a device.

The units used are volts (V), plus their derivatives: megavolts (MV), kilovolts (kV), millivolts (mV) and microvolts (μV).

The unit of 1 V is defined as the potential difference across a conductor when a current of 1 amp dissipates 1 W of power (BIPM 2006).

In radiology, a similar sounding but separate unit, the electron volt (eV), may be used to describe the energy of electrons, particles, x- and gamma-rays.

**Related Article:** Electron volt

**Further Reading:** BIPM. 2006. The International System of Units (SI), Stedi Media, France.

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**Voltage divider**

(*)General*) The voltage divider refers to a simple electrical circuit made up of two impedances (AC) or two conductors (DC).

The major property of the voltage divider is that in the following layout, the output voltage is directly related to the input voltage by the ratio of impedances (Figure V.12).

Resistors are commonly used to provide a voltage divider as these will precisely divide both DC and AC signals of all frequencies.

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**Example:** To explain voltage drop, we shall use a very simplified example. If an exposure is planned to be made with 100kVp, 100mA, this will require energy of the order of 10 kW. For a single phase 200 V mains, this will produce a current pulse (during the exposure) of the order of 50 A (10 kW/200 V = 50 A).

If we assume that the resistance of the mains R is 0.2 ohm (what is a very good low resistance), then the voltage drop due to the R only will be 10 V (50 A × 0.2 ohm = 10 V). Having on mind that the HV transformer coefficient is approx. 500, these 10 V drop in the primary side of the HV transformer will produce approx. 5 kV drop of the high voltage to the tube (i.e. 95 kV instead the planned 100 kV). This will lead to change of the planned x-ray spectrum and hence the image contrast.
In order to correct this voltage drop, the autotransformer has to supply the HV transformer with 10 V more (i.e. 210 V). As a result, the 10 V voltage drop in the mains during the exposure will be compensated (210 − 10 = 200 V), and the x-ray generator will deliver exact exposure of 100 kV and 100 mA (only during the first millisecond, there will be a short kV jump to 105 kV).

**Related Articles:** Anode, High-voltage generator, High-voltage transformer

**Voltage limiter**

(General) A voltage limiter acts on an electrical signal to prevent it exceeding a preset maximum and/or minimum value, limiting the signal to a specific range of values.

Many electronic circuits are susceptible static electricity and signal exceeding their power supply values, so a voltage limiter circuit is commonly found at their input to avoid damage and ensure safe operation (Figure V.14).

Commonly used circuits shown in the following are designed to ‘shunt’ any excessive signal to ground whilst containing a resistor to prevent excessive current and protect both source and limiter circuits (Figures V.15 and V.16).

**Related Articles:** Diode, zener, Static electricity, Suppressing filter

**Voltage range**

(General) The voltage range of a circuit represents that range of voltages over which the input (or output) of a device can function correctly.

All electronic components will have specified operating limits, outside which the devices will not function properly and safely. As a complete unit, an electronic circuit will not usually operate correctly when the input voltages exceed the power supply values, and the outputs are rarely capable of providing signals, which swing as far as the power supply voltages.

Should circuits be driven past their operating voltage range, their outputs usually ‘limit’ if designed well, but can on occasion swing violently in value or even latch up.

**Related Article:** Voltage limiter

**FIGURE V.14** Voltage limiter principle.

**FIGURE V.15** Diode voltage limiter maintains signal within +/- supply range.

**FIGURE V.16** Zener voltage limiter limits signal to +/- Zener values.

**FIGURE V.17** Voltage regulation.

**Voltage regulation**

(General) Voltage regulation refers to the process by which a voltage source (usually a DC power supply) is regulated or maintained at some constant value when subject to varying input and output conditions.

A power source may be subject to varying input voltages and will certainly be subject to changing current demand by the circuitry it feeds. Some form of regulation is therefore required to make the supply maintain its output voltage as constant as possible. Electrical noise, mains ripple, etc. should also be minimised by the regulator circuitry.

Voltage regulation is performed by either shunt or series regulation, though except for the zener shunt regulator, nearly all are of the series regulation type, where some form of feedback or servo mechanism monitors the output voltage, compares it with some internal standard voltage and automatically reacts to maintain the output voltage constant (Figure V.17).

More advanced regulators incorporate over-temperature shutdown, short-circuit protection and current limiting, which help to protect itself and the circuitry it serves. However, this can cause problems and it may be necessary to test the regulator with a ‘dummy load’ to ensure it is functional.

**Related Articles:** Zener, Voltage stabiliser

**Voltage stabiliser**

(General) See Stabilisation

**Voltage supply**

(General) A voltage supply is the name given to a source of electrical power, which maintains its voltage whilst providing a varying current as required by the equipment or load attached.

Two types of voltage supply are frequently referred to: the mains voltage supply and the DC power supply.

**Mains Voltage Supply:** The ‘mains supply’ is an AC power source with properties and limits specified nationally. Common domestic and office supplies are single phase, 50 or 60 Hz frequency and 110–230 V rms depending on country, and the maximum current available from each outlet is limited (to protect wiring etc.) to the range of 0–20 amp, again depending on country.

Higher power (greater than about 3 kW) is usually only available where higher-rated cabling has been installed in the building, and nonstandard plug/socket arrangements are used.

For high powers such as those required for CT, MR scanners and some x-ray sets, ‘three-phase’ supplies are used again specially
Cabled to the required equipment and usually hard wired to equipment via mechanical circuit breakers.

**DC Power Supply:** DC power supplies are common in all mains powered equipment and are nearly always ‘voltage supplies’ – that is, they maintain their output at a constant voltage, independent of changed input voltage and output current demand.

**Related Articles:** Mains supply circuit, Mains voltage, Voltage regulation

**Voltage waveform**

*Diagnostic Radiology* The voltage waveform of an x-ray generator is the variation of the kV value during the exposure. This is directly related to the type of x-ray generator (e.g. the rectification in classical x-ray generator; the frequency of medium frequency x-ray generator, etc.).

These kV fluctuations are also called kVp ripple. The ripple factor represents the maximal variation of the kVp during the exposure. The voltage waveform is measured through oscilloscope either directly (using high-voltage divider, connected directly to the generator) or indirectly (using special kVp meter).

The kV voltage waveform is an important factor for the formation of the x-ray spectrum. The more and the higher are the kVp fluctuations, the greater is the proportion of the generated low energy x-rays (produced by the low kV components of the waveform). These low energy photons are anyhow absorbed by the tube housing filtration; hence the effectiveness of the x-ray tube is decreased. Contemporary medium frequency x-ray generators produce smooth kV voltage waveform (with minimal ripple).

Figures V.18 through V.23 present some typical kV voltage waveforms (all these are measured with Keithely kVp meter). Note

**FIGURE V.18** kVp voltage waveform of single-pulse classical x-ray generator. y-axis – kVp; x-axis exposure time (1 division = 20 ms).

**FIGURE V.19** kVp voltage waveform of single-pulse dental classical x-ray generator. Note the initial pulses for pre-heating of the cathode with gradually increasing kV amplitude (1 division = 50 ms).

**FIGURE V.20** kVp waveform for two pulse classical x-ray generator – Below (1 division = 20 ms). The exposure time is 118 ms. The pulsations are 100%. The curve above the kVp waveform is the dose output waveform – It follows the kVp waveform.

**FIGURE V.21** kVp waveform for six-pulse classical x-ray generator – Below (1 division = 5 ms). The exposure time is 22.1 ms. The pulsations are approx. 14%. The curve above the kVp waveform is the dose output waveform.
voltages (kV) can reliably be measured this way. This device does have a great advantage that it takes no current and so does not load the circuit being measured.

The electromagnetic voltmeter relies on the applied voltage generating a proportional current in a small moving coil, which twists in reaction to a fixed magnetic field (DC meter) or attracts a piece of soft iron on a spring mount (AC and DC). Meter movements capable of full scale deflection with a current of only about 50 μA are common, and where this amount of current will not affect the real value of the circuit under test, this sort of movement with an appropriate series resistor provides an excellent method of displaying voltage. The quality of these meters is defined by the ‘ohms/volt’ value of the movement. A value in the region of 50 kΩ makes for a good general-purpose device.

Digital voltmeters and multi-meters aim to mimic and improve on their analogue counterparts both in the accuracy possible by multi-digit displays and in their ability to operate whilst taking much smaller currents and hence having less effect on the circuit under test. This is possible by incorporating high-input-resistance amplifiers preceding the analogue-to-digital converters and display. Inexpensive multi-meters can be found with input impedances in the megohm region.

Specialist meters, which can take virtually no current, ‘electrometer’, are also available for measurement where the voltage to be determined is in a circuit where to take even a small current would affect the circuit under test. These are often used to measure bio-potentials directly, such as from individual living cells.

**Volume of interest (VOI)**

*(Nuclear Medicine)* In planar scans, images are often evaluated by means of region of interests (ROI) that is defined over the organ of interest using a pointing device. The area of the ROI is then determined by the number of pixels inside the ROI times the square of the pixel size. A three-dimensional (3D) image consists of a set of digital images, where each image is defined by a matrix of numbers that define some estimate in a location corresponding to the pixel location in the image. For example, a pixel value in a digital image obtained from a SPECT scan reflects the activity concentration in that particular location. Usually, one defines a ROI using a track-ball or mouse that calculates, for example the sum of pixel values within a boundary, for example the kidney. Since each pixel defines an area and each image also defines a slice thickness, in the 3D space, this will be a voxel (volume element). By defining ROIs in consecutive slices, a volume called VOI or volume-of-interest is obtained. The volume then will be

\[
\text{Pixel size } X \times \text{Pixel size } Y \times \text{Slice thickness } Z
\]

**Abbreviations:** ROI = Region of interest, VOI = Volume of interest and 2D = Two-dimensional

**Volumetric-intensity-modulated arc therapy**

*(Radiotherapy)* Volumetric-intensity-modulated arc therapy (VIMAT sometimes known as IMAT) or volumetric-modulated arc therapy (VMAT) is an advanced form of intensity-modulated radiotherapy (IMRT) technique. The technique is based on a concept proposed by Cedric Yu.

Unlike step and shoot or dynamic MLC IMRT treatment techniques, the gantry angle in a VMAT treatment varies continuously as in arc therapy, and, at the same time, the intensity of the treatment beam is modulated with dynamic MLC motion. The technique can deliver highly conformal treatment faster and with less machine monitor units than that as given by conventional IMRT.
but with similar dose conformality. The VIMAT treatment mode is now commercially available and has been clinically implemented for routine clinical use.


**Volumetric prescribing (brachytherapy)**

*(Radiotherapy, Brachytherapy)*

**3D Treatment Planning in Brachytherapy:** Three-dimensional images are used for both external beam and brachytherapy treatment planning, and acronyms like IGBT – image-guided brachytherapy – and IGABT – image-guided adaptive brachytherapy – have been used in the brachytherapy community.

With target volumes and organs at risk delineated, tools like dose volumes histograms are also available to evaluate dose distributions in brachytherapy.

**ICRU Recommendations – Dose and Volume Specification for Reporting:** ICRU has published two brachytherapy reports; the ICRU Report 38: Dose and Volume Specification for Reporting Intracavitary Therapy in Gynaecology (1985), and the ICRU Report 58: Dose and Volume Specification for Reporting Interstitial Therapy (1997). ICRU also recommends that the same terms and concepts are used in brachytherapy as in external beam radiotherapy, that is a consistent language used whenever possible.

ICRU 58 also states that it is not the intention to ‘encourage users to depart from their normal practice of brachytherapy and dose prescription. The aim is to develop a common language which is based on existing concepts. It should be usable to describe what has been done in a way that can be more closely related to the outcome of treatment and one that is generally understood’.

**Developments:** With the development of 3D treatment planning, there are also developments in the recommendations on reporting brachytherapy, both for cancer of the cervix and for prostate cancer. These new recommendations, both for intracavitary and interstitial brachytherapy, use dose-volume-histogram information to evaluate dose distributions and relate dose-volume histogram parameters to treatment results (see Salambier et al. and Kovács et al. for prostate brachytherapy and Haie-Meder et al., Pötter et al. and Lang et al. for cervix cancer brachytherapy).

**Volumetric Prescription:** In the abstract of Haie-Meder et al., the authors state (cervix cancer brachytherapy), ‘It is expected that the therapeutic ratio including target coverage and sparing of organs at risk can be significantly improved, if radiation dose is prescribed to a 3D image-based CTV taking into account dose volume constraints for OAR. However, prospective use of these recommendations in the clinical context is warranted, to further improve and develop the potential of image-based cervix cancer brachytherapy’.

**Abbreviations:** CTV = Clinical target volume, ICRU = International Commission on Radiation Units and Measurements and OAR = Organ at risk

**Related Articles:** Dose volume histograms – Brachytherapy, Image-guided brachytherapy


**Voxel**

*(Nuclear Medicine)* A volume element in three dimensions. A voxel is the 3D analogue of the 2D pixel. Voxels are used to visualise 3D medical data. Typically, a voxel contains no information regarding its position in space. The position is derived from the position relative to the other voxels in the data file.
Wall filter
(Ultrasound) This is a high-pass filter that blocks low-frequency components of the Doppler signal. In Doppler measurements, the Doppler shift frequency is detected. If however the target is stationary, a DC-signal is received. This corresponds to the phase shift between the transmitted and received signals, which is constant as the target is stationary. As tissue has on the order of 10–100 times higher scattering strength than blood, a high-pass filter must be inserted in order to eliminate this DC-component. Otherwise, the lower part of the power spectrum may be distorted due to windowing and segmentation of the data. High DC-components can also be a problem in the signal processing due to clipping and thereby distorted results.

The filter limits the lowest velocity that can be estimated, and the limit has to be a trade-off between the desired lower velocity limit and tissue movements, transducer movements and settling times for the filter. Moreover, as the blood signal is so much weaker than the tissue signal, the filter order has to be considered so that there is sufficient suppression of the low-frequency tissue signal for the chosen –3 dB limit.

Warm-up
(Diagnostic Radiology) Warming up of the x-ray tube is needed for two main reasons. First, the anode of a cool x-ray tube has to be slowly heated, as very quick heating can lead to thermal stress and cracks on its surface. Such warm-up can be performed at the beginning of the working day (often done for CT scanners) or before an examination with heavy exposures (often done before QC tests).

Second, warm-up may be needed if the tube has not been used for long period of time (several months or more). In this case, it is possible that the x-ray tube vacuum has degreased (due to internal ionisation), and a heavy exposure can lead to internal arcing and destruction of the x-ray tube. In this case, the warm-up uses exposures with gradually increasing power. These slowly consume the occasional ions in the tube, this way restoring the vacuum (degassing of the x-ray tube).

As an example for x-ray tube warm-up, at least three warming exposures have to be performed:

1. Approximately 50kV, 50mA and 100 ms
2. Approximately 50kV, 100mA and 200 ms
3. Approximately 70kV, 100mA and 200 ms

The minimum interval between the exposures has to be at least 1 min.

Related Articles: Anode, Cooling curve, Thermal stress

Warm-up time
(Diagnostic Radiology) See Warm-up

Warning lights
(Radiation Protection) Warning lights should be used to clearly indicate when equipment that emits ionising radiation is functioning in one room of the controlled areas. A green/red light above the door to the room housing the equipment should be green when it is possible to enter the room because the equipment is ‘off’ and shall turn red when the equipment is ‘on’, that is emitting radiation. To see the red light should in fact prevent people from accidentally entering the room. Standard warning lights should be used as recommended by the international organisations. Depending on the nature of the equipment and its applications, different systems shall be used. As an example, in radiotherapy, the system should be such that a teletherapy unit must cease to operate when the entrance door is opened (there will also be other safety devices in addition to the light). On the other hand, in a diagnostic fluoroscopy room, the red light should only be a warning in order to avoid unnecessary entrance, but opening the door shall not turn off the equipment.

Related Articles: Controlled area, Supervised area, Warning sign


Warning sign
(Radiation Protection) Warning signs shall be placed to clearly indicate the demarcation of controlled and supervised areas. The warning signs to be used are those recommended by the International Organization for Standardization (ISO). The signs also give the possibility to indicate the nature of the radiation and therefore the related danger: only external irradiation (as, e.g. with x-ray diagnostic equipment) or internal contamination or both (as, e.g. in nuclear medicine).

Related Articles: Controlled area, Supervised area


Washing in film processing
(Diagnostic Radiology) After going through the fixer solution film is next passed through a water bath to wash the fixer solution out of the emulsion. It is especially important to remove the thiosulphate. If thiosulphate (hypo) is retained in the emulsion, it will eventually react with the silver nitrate and air to form silver sulphate, a yellowish brown stain. The amount of thiosulphate retained in the emulsion determines the useful lifetime of a processed film. The American National Standard Institute recommends a maximum retention of 30μg/in.².

Waste disposal
(Radiation Protection) For the general definition, see Radioactive waste. In medical applications, it is often feasible to organise locally a system for waste disposal. Depending on the characteristics of the radioisotopes used, it is important to evaluate the possibility to store and sort out waste substances until the level of radioactivity is below the limit allowed for general waste. In this case, it might
also be possible to re-use some material, for example the lead in the molybdenum generators used in nuclear medicine. In case where local organisation of the waste disposal would not be possible, it is advisable to make contracts with external companies (usually the same organisation that supplies the radioisotopes).

Related Article: Radioactive waste

Waste, radioactive
(Radiation Protection) See Radioactive waste

Water
(General)

<table>
<thead>
<tr>
<th>Property</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Molar mass</td>
<td>18.015 g/mol</td>
</tr>
<tr>
<td>Density at STP</td>
<td>998 kg/m³</td>
</tr>
<tr>
<td>Melting point</td>
<td>273 K</td>
</tr>
<tr>
<td>Boiling point</td>
<td>373 K</td>
</tr>
<tr>
<td>CT number</td>
<td>0 HU</td>
</tr>
</tbody>
</table>

Water is a common substance that covers 71% of the Earth’s surface, which is essential for the existence of life. At room temperature and pressure, it is in liquid form, but it is commonly found in all three states, known as ice when solid and steam when gaseous. Water exists in nature mostly in the oceans (97%), rivers, aquifers, air, clouds, precipitation and glaciers. Water is also contained within biological organisms. It moves continually through a hydrologic cycle of evaporation/transpiration, precipitation and runoff to the sea. Although fresh water is essential for life, there is an increasing shortage in many parts of the world.

Pure water is a tasteless, odourless and has a very light blue colour. However, water readily dissolves many substances, giving it various tastes and odours. Water is miscible with many substances, except most oils. It is transparent, only absorbing strong UV light. Water is a polar molecule (Figure W.1) due to the dipole moment created by the oxygen and hydrogen atoms. This causes the molecules to attract strongly via hydrogen bonding giving water one of the highest specific heat capacities, allowing it to moderate the Earth’s climate. The van der Waals interactions between the molecules give water its characteristic property of high surface tension. Water exhibits capillary action, a tendency to move up a narrow tubes against gravity, which is vital in all vascular plants. Water is a strong universal solvent, essential for the transport of many substances in cells, such as DNA. Pure water has a low electrical conductivity, but this increases considerably with ionic solutes, such as sodium chloride. Water is unique in that it becomes less dense upon freezing, causing ice to float on water.

Water is significant in the world economy, as it is extremely versatile in its applications. It is used most extensively in agriculture for the irrigation of crops. Water is vital for drinking, as the human body is 55%–78% water and requires several litres a day to function properly. It is useful as a solvent in industry, for cleaning and for the transportation of sewage waste. It is used for heat transfer for both heating and cooling, such as for driving steam turbines in electric power plants. Water is effective at extinguishing fires (excluding electric and chemical fires) as it has a high heat of vaporisation and is relatively inert. Water is also extensively applied in the chemical industry, power generation (hydroelectricity) and food processing.

In a nuclear power plant, the reactor core is immersed in a suitable coolant. Fission occurs in the nuclear fuel, and the fission energy in the form of kinetic energy of fission fragments and new neutrons is rapidly converted into heat. The coolant (usually water) is used to maintain a stable temperature in the reactor core and exits the core either as steam or as hot pressurised water, subsequently used to drive turbines connected to electric power generators.

In a nuclear reactor, moderators are used to slow down the newly produced fast neutrons through elastic scattering events between neutrons and the nuclei of the moderator. Water serves as a moderator material in most reactors; however, some reactors may use the so-called heavy water (deuterium based), graphite or beryllium for the purposes of moderation. Heavy water has a smaller probability for neutron absorption through the (n,γ) reaction than water; however, it is much costlier.

Medical Applications: Water is applied widely in medicine, for example for cleaning and as saline for intravenous use. It is often used as a scientific standard, such as for the Kelvin and Celsius temperature scales and also the Hounsfield scale to quantify a material’s radiodensity in relation to water. Properties of human soft tissue are often approximated to that of water, due to its high water content. This makes water an effective and readily available phantom material for medical physics applications.

Related Articles: CT Number, De-ionised water, Solid water phantom, Water calorimeter, Water cooling, Water suppression, Water tank

Water calorimeter
(Radiation Protection) The absolute measurement of absorbed dose in water can be performed using a sealed water calorimeter. The amount of heat ΔQ produced in the water by ionising radiation is equal to

\[ \Delta Q = c_w \times m \times \Delta T \]

where

- \( c_w \) is the water specific heat
- \( m \) is the mass of water
- \( \Delta T \) is a change of temperature

The measurement of \( \Delta T \) enables us to evaluate the absolute amount of energy absorbed by the water. The absorbed dose \( D \) is defined as energy absorbed per unit mass. The value of \( D \) can be calculated from the equation

\[ \Delta D = c_w \times \Delta T \]

Related Articles: Calorimeter, Calorimetry


FIGURE W.1 Chemical structure of water (H₂O) showing the bond angle and bond length.
Water cooling

(Diagnostic Radiology) X-ray tube with direct water cooling of the anode is rarely used these days. Such tube requires the anode to be grounded (0V) and the cathode bears all the voltage (e.g. −100 kV). In this case, the stem of the anode has special construction, allowing running water to pass through and quickly cool it. However, most often, the anode bears high positive voltage and is cooled by insulating oil, which oil is in turn cooled by running water (i.e. indirect water cooling).

Related Articles: Anode, Stationary anode, Rotation anode, Cooling curve, X-ray tube housing

Water cooling

(Magnetic Resonance) Gradient systems dissipate a lot of heat due to the high current loads and heavy duty cycle. This can have a detrimental effect on the magnet resonance imaging (MRI) system. Gradient coils are therefore equipped with closed-loop water-cooling systems.

Resistive magnets require large currents to generate the static magnetic field, and significant water cooling of the magnet coils is required for such MRI units.

Related Articles: Duty cycle, Gradient coil, Magnet resistive

Water suppression

(Magnetic Resonance) A conventional MR image is essentially a map of water distribution (spin density), and MRI would not be feasible at all, on the grounds of spatial resolution and imaging speed, if it was not for the large amount of water present in the body. In proton magnetic resonance spectroscopy (MRS), however, the desire is to detect proton-containing compounds (e.g. metabolites) at much lower concentration. Because the concentration of water in body tissues is around 10,000 times that of metabolites, unless special measures are taken the water peak dominates any in vivo proton spectrum to the extent that other signals are buried beneath it and occupy a tiny proportion of the dynamic range of the analogue to digital converter. This problem delayed the development of in vivo proton spectroscopy until suitable water suppression techniques became available, a process that required pulse sequence development and in many cases also hardware improvements that took some years to achieve.

Today, a range of water suppression techniques is available, and clinical MRI systems usually allow the user to choose between several methods.

Frequency-Selective Excitation and Refocusing: Frequency selective RF-pulses can be used to selectively excite the frequency range of metabolites whereas the water resonance remains unaffected. An alternative approach, frequency selective refocusing, employs a spin-echo sequence designed to refocus only the metabolite magnetisation, while dephasing that of water.

Frequency-Selective Saturation: A widespread approach to water suppression in the context of in vivo MRS is the use of one or more frequency selective RF pulses, designed to tip water magnetisation into the transverse plane, followed by spoiler gradient to dephase this magnetisation. Elimination of the water signal in this way is followed by a conventional MRS sequence, which detects only the remaining metabolite signals.

The most common implementation of this approach is the CHESS (chemical shift selective) sequence of Haase et al. In practice, selective excitation is generally repeated two or three times to improve water suppression.

Composite and Binomial Pulses: Both water suppression techniques, frequency selective excitation and saturation require frequency selective RF-pulses. Composite and binomial pulses are often applied for this purpose.

A composite pulse is a sequence of RF pulses designed to emulate the behaviour of a simpler pulse, but with special features such as tailored off-resonance behaviour or insensitivity to magnetic field inhomogeneity. In the context of water suppression, composite pulses can be designed to generate a very small flip angle at the resonance frequency of water, so that very little water signal is collected relative to metabolites at other frequencies.

Binomial pulses are trains of RF pulses, with the flip angles generated by successive pulses given by binomial coefficients. For example, the ratios of the amplitudes of the pulses in the three simplest binomial sequences are 1:1, 1:2:1 and 1:3:3:1. Phase alternative between successive pulses may also be used. The usual approach is to place water on resonance and choose the intervals between pulses, with respect to the frequencies of the metabolites of interest, so that off-resonance effects ensure that the sequence as a whole places magnetisation from these metabolites into the xy-plane, while water signal is returned to the z-axis and produces no signal. It is possible to tailor the width of the spectral region in which transverse magnetisation is generated and that of the null region, by using sequences of increasing complexity.


Water tank

(Radiotherapy) A water tank is a large Perspex tank (typically at least 40 × 40 × 40 cm³), which is filled with water for dosimetric work. An arm system allows a range of detectors to be positioned very precisely throughout the volume to acquire detailed information such as beam profiles and percentage depth doses. Typically, the water tank is used mainly at acceptance testing and commissioning, or whenever a major modification is made to the linac. The time of set-up renders the water tank less useful for routine QC measurements, and so solid water materials are used for these measurements.

Abbreviation: QC = Quality control.

Related Article: Solid water phantom

Watt

(General) The watt (symbol W) is the Système International (SI) unit of power (i.e. the rate of expenditure or transformation of energy from one form to another). Therefore, in SI units, 1 W = 1 J/s.

Similarly to joule, watt is a derived SI unit and may more properly be expressed in terms of base SI units as

\[1 W = \frac{kg \cdot m^2}{s^3}\]

where

- kg is kilogram (mass)
- m is metre (distance)
- s is second (time)

The unit is named after James Watt, Scottish inventor and mechanical engineer (1736–1819).

Related Article: Système International


Wave equation

(Ultrasound) Any parameter \( \Phi \), being a function of time and position, that satisfies the equation

\[ \frac{\partial^2 \Phi}{\partial t^2} = c^2 \frac{\partial^2 \Phi}{\partial x^2} \]
can propagate as a wave at speed c in the positive or negative direction. These functions generally have arguments of the type $(ct - x)$ or $(ct + x)$ respectively.

We can regard pressure as such a parameter, which propagate due to the elasticity and density of the medium carrying the sound wave. The elasticity (or compressibility) makes the medium return to the state it had before the disturbance occurred. The mass of the medium (density) will give inertia to this returning motion so that the movement will have an overshoot. If then the elements of the medium are coupled, a wave can propagate. To describe this wave motion mathematically, one studies the small deviations from equilibrium pressure: a small deviation in pressure will cause a corresponding deviation in density. Three equations (or relations) are needed: (1) the conservation of mass (the change of mass in a volume must equal the difference between the mass entering and leaving the volume), (2) conservation of momentum (an equation of motion), and (3) that density can be expressed as a function of pressure only (an equation of state). Relation (3) will be true under the assumption that the process is adiabatic, that is that no heat transfer occurs between compressed and rarefacted regions. By combining these three statements, one arrives at the well-known wave equation given earlier. The derivation is based on small variations in density of the medium caused by the pressure variations in the sound wave. The result is valid for a region with homogeneous density and compressibility.

**Waveguide**

(Radiotherapy) A waveguide is an evacuated or gas-filled tube structure that guides waves, transferring information or power. They are used in linear accelerators to accelerate the electrons to high energy to form the therapy beam in radiotherapy (accelerating waveguide) as well as coupling the microwave energy from the klystron or magnetron source (power transmission waveguide).

**Accelerating Waveguide:** There are two main types of accelerating waveguide: standing wave or travelling wave, which refer to the type of electromagnetic wave that is established within the guide. The phase velocity of the electromagnetic wave can be adjusted by placing internal iris diaphragms within the circular guide, creating internal cavities that form the basic structure. The electrons are injected into one end of the waveguide and ‘ride’ upon the crests of the electromagnetic wave. By increasing the phase velocity of the electromagnetic wave, the electrons can be accelerated up to MeV energies.

- **Travelling wave:** In this type of waveguide, the electromagnetic wave enters the tube and propagates down towards the end, where it is either absorbed or exits to be fed back into the input end. There is no reflection of energy. These waveguides generally have two sections. In the first ‘buncher’ section, the cavities are non-uniform (inner diameter, aperture diameter and axial spacing vary). This section groups the electrons together so they are all moving coherently in space, phase and velocity. The last section is the ‘accelerator’ section, which transfers energy to the electrons until they reach MeV energies. This structure can be seen in Figure W.2.
  - **Standing wave:** In contrast, a standing wave structure has conductive discs placed at each end that reflects the electromagnetic power, so a standing wave is built up. A schematic of the standing waveguide is shown in Figure W.3.

The arrows show the axial electric field at an instant in time. An electron in cavity 1 will receive acceleration at that instant. The standing wave will then reverse the direction of the axial field at the oscillation frequency. If the speed of the electron is such that it arrives in cavity 3 when the standing wave has reversed, then it will receive acceleration again at that point. The electron injections are hence synchronised with the frequency of the RF power source so this occurs. Cavities 2, 4 and 6 are positioned at the nodes of the standing wave and hence never transfer energy to electrons. This fact can be exploited to shorten the tube, by moving these cavities to the sides – a side-coupled waveguide. A cut-away view can be seen in Figure W.4. This has advantages in that it produces a higher accelerating gradient per metre, and the reduced length of the waveguide can be mounted directly in the gantry. However, they do also have greater lateral bulk, which forces everything further away from isocentre, hence needing a wider turning circle and higher patient table.

**Power Transmission Waveguide:** Rectangular waveguides are used to transfer the microwave energy from the source to the accelerating waveguide. A circulator (or isolator) must be placed in each section to isolate the source from the cavity. A schematic is shown in Figure W.5.
between the power source and the waveguide to protect the source from any reflected power travelling in the opposite direction. This allows radiation to pass through that is travelling from the source, but blocks radiation travelling to the source.

Related Article: Linac

Wavelength
(Ultrasound) Wavelength is the shortest distance between two points that are in phase on a sinusoidal wave, that is the distance of a complete cycle of the wave. The wavelength, \( \lambda \), is related to the frequency by \( \lambda = c/f \), where \( c \) is the wave propagation speed (or strictly phase speed, that of an infinitely long sinusoidal wave). Thus, it is somewhat difficult to talk about the wavelength for a broadband (i.e. short) pulse.

Wax
(Radiotherapy) Wax can be used as bolus, which is a tissue-equivalent material placed directly on the skin surface to even out the irregular patient contour and thereby provide a flat surface for normal beam incidence. See article Bolus.

Related Article: Bolus

Weber
(Magnetic Resonance) The Weber, abbreviated Wb, is the SI unit of magnetic flux and is named after the German physicist Wilhelm Eduard Weber (1804–1893).

The Weber is a large unit that is described as the amount of magnetic flux that in 1 s produces 1 volt per turn of a linked circuit.

Expressed in SI base units, the Weber is written as

\[
\frac{\text{kg} \cdot \text{m}^2}{\text{s}^2 \cdot \text{A}}
\]

which is also equal to

\[
\text{T} \cdot \text{m}^2
\]

Related Article: Flux

Wedge
(Radiotherapy) A wedge (or wedge filter, compensating wedge) is an external compensator that can be physically placed in the beam to create oblique dose profiles across the central axis of a beam, due to its tapering thickness in one dimension. They are used to either compensate for an oblique surface (illustrated in Example A) or to correct for non-uniformities in multi-beam plans (illustrated in Example B).

Figure W.5 shows a typical isodose distribution obtained from a wedge filter. The beam is attenuated greater at the thicker end of the wedge (the heel), leading to a higher dose at the thinner end of the wedge (the toe).

Three types of wedge filters are currently in use: physical, motorised and dynamic.

A *physical wedge* is an angled piece of material (lead, brass or steel) that is placed in the beam to produce a gradient in radiation intensity. Manual intervention is required to place physical wedges on the treatment unit’s collimator assembly. Physical wedge filters are widely used in both cobalt units and linear accelerators. They are energy-specific. The wedge angles are limited to those available, often 15°, 30°, 45° and 60°. However, intermediate angles can be effectively created over the whole treatment by swapping to an alternative wedge midway through treatment.

A *motorised wedge* is a similar device, a physical wedge, typically 60°, integrated into the head of the unit and controlled remotely. Any wedge angle can then be created by combining the wedged field with an open field of varying duration. The advantages of this method over physical external wedges include greater flexibility, reduced manual handling risks for the radiographer and reduced set-up time for the patient. This technique is offered in Elekta linacs.

A *dynamic wedge* produces a wedged intensity gradient by moving one jaw across the field whilst keeping the other stationary, which creates a wedged field since different parts of the field are
irradiated for different times. The advantage of this method is that the beam does not undergo any beam hardening that would otherwise be associated with a physically wedged field. Dynamic wedges are offered for Varian linacs (enhanced dynamic wedge [EDW]) and Siemens linacs (virtual wedge [VW]). See article on Dynamic wedge for further information.

Example A: Wedge filters are routinely used in irradiation of the breast: two wedged (\(\sim 15^\circ\)) coaxially opposed beams are used to compensate for the oblique incidence (Figure W.6).

Example B: A three-field plan is often used for irradiation of the pelvis, with two laterally opposed wedged fields and an anterior field. The wedges are necessary to balance the dose distribution within the target volume (Figure W.7).

Related Articles: Wedge angle, Wedge transmission factor, Dynamic wedge, Beam hardening


Wedge angle

(Radiotherapy) The wedge angle of a wedged field can be defined in two ways. The IEC standard defines the wedge angle as that of the isodose curve to a plane perpendicular to the central axis, at a depth of 10 cm. Alternatively, it is the angle that the 50% isodose line makes with the central axis of the beam.

The wedge angle required to act as a tissue compensator is normally between 50% and 75% of the angle of the surface obliquity depending on depth and energy.

For treatments that combine the use of two wedges, the effective wedge angle of the combined fields is the weighted average of the tangents of the two wedges, weighted according to the relative contribution to the dose at the depth of dose maximum:

\[
\tan \theta_0 = \frac{w_1 \tan \theta_1 + w_2 \tan \theta_2}{w_1 + w_2}
\]

Wedge angle calculation, where \(w_1\) and \(w_2\) are weighting factors.

For example, the motorised wedge uses a combination of a 60° wedge and an open field. If equal weighting is given to each field, then the effective wedge angle is given by

\[
\theta_0 = \arctan(\tan(60^\circ)/2) = 41^\circ
\]

Related Articles: Wedge filter, Wedge transmission factor, Dynamic wedge, Tissue compensation, Wedge

Wedge field

(Radiotherapy) See Wedge

Wedge filter

(Radiotherapy) See Wedge

Wedge transmission factor

(Radiotherapy) Wedge transmission factor (wedge factor, WF) is the ratio of the dose with the wedge in the beam to the dose in the same conditions but without the wedge in the beam. This is generally a function of both depth and field size and hence is normally measured at a specified depth on the central axis for the reference field size. Wedge factor measurements should be included in routine QA checks. When the wedge factor is measured, care must be taken to place the ion chamber axis parallel to a constant thickness of the wedge. The wedge factor is often compared to the opposing wedge factor (by rotating wedge or collimator by 180°) to ensure that the position of the ion chamber/wedge is correct. Rotation of the collimator verifies that the ionisation chamber is positioned on the collimator axis of rotation. Rotation of the wedge itself reveals whether the side rails are symmetrically positioned about the collimator axis of rotation. The measured values should be within tolerance limit for the two wedge orientations. Usually, the average value of the two wedge orientations is taken as the correct value of the wedge transmission factor.

Abbreviation: WF = Wedge factor.
Related Articles: Wedge filter, Wedge angle, Wedge


Wehnelt electrode

(Diagnostic Radiology) Normally the beam of thermal electrons produced by the cathode filament is quite spread, resulting in an increased...
area of the focal spot. This enlarged size of the source of radiation blurs the x-ray image. In order to focus the beam of thermal electrons and to decrease the space-charge effect, the cathode filament is placed in a focusing cup (a half-pipe groove, known also as Wehnelt electrode or Wehnelt cylinder, named after its inventor). The focusing cup is specially shaped and is made of molybdenum, nickel or steel, because of their poor thermionic emission. The cup can be equipotential with the cathode. In this case, placing the filament wire inner or outer in the cup changes the focusing of the electron beam (Figures W.8 and W.9).

The cup can also be charged with slight negative potential against the cathode (of the order of tenths of volts). Changing the negative charge of the Wehnelt electrode leads to change in the focusing effect (Figure W.10). In this way, the electron flow (anode current) can be controlled (even stopped). Such x-ray tubes (known as grid-controlled) allow the creation of very short x-ray pulses (of the order of 1 ms), which are especially useful for imaging of dynamic objects (e.g. the heart). Switching on/off the x-ray beam through the grid-control is technologically easier and above all far quicker than switching on/off the high anode voltage.

**Related Articles:** Cathode, Focal spot, Space-charge effect, Grid-controlled tube

**Hyperlink:** EMERALD (DR module), www.emerald2.eu

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**Well-counter detector**

*(Nuclear Medicine)* A counter system that consists of a single crystal including a hole within it for insertion of radioactive substances. The aim of the design is to surround the substance with as much detector as possible (approaching a solid-angle of 4π), increasing the efficiency of detection and pushing down the minimum detectable activity (MDA). These simple detectors are used to estimate the activity in a radioactive sample.

The common well-counter design consists of a single crystal attached to a PM-tube sealed by a high attenuating material (lead shielding) as seen in Figure W.11. The crystal typically used for well-counter detectors is NaI (Tl). Lead shielding is employed to reduce the background radiation contribution to the count rate.


**Well-counter detector efficiency**

*(Nuclear Medicine)* The ratio of the number of detected photons to the total number of emitted photons from a sample is referred to as...
to as the well-counter detector efficiency. If a large fraction of the emitted photons is detected and registered by the well-counter, the detector efficiency is close to 1. The total efficiency is the product of the intrinsic and geometric efficiency.

The detection efficiency of a well-counter for most typical $\gamma$-emitters is high. The primary reason is the high geometric efficiency $g$. A sample in a well-counter is almost entirely surrounded by the crystal hence most of the emitted photons pass through the crystal. The intrinsic efficiency $\varepsilon$ depends on crystal thickness and $\gamma$-ray energy because a thick crystal increases the probability of photon attenuation in the same way as higher energy leads to higher photon penetration. The calculation of the intrinsic efficiency is complicated because the crystal thickness is different for different emission angles, that is a photon with a perpendicular angle of incidence has less absorptive crystal to pass through than a photon with an oblique angle of incidence.

In some well-counters, counts outside the photopeak are discriminated, which makes it necessary to account for the photon fraction $f_p$ when calculating the efficiency. The photofraction is inversely proportional to the $\gamma$-ray energy and proportional to the crystal thickness. The intrinsic photopeak efficiency $\varepsilon_p$ is the product of photofraction and the intrinsic efficiency, and it is a more appropriate measure of the efficiency in a photopeak-only well-counter system:

$$\varepsilon_p = \varepsilon \times f_p$$

(W.1)

Related Article: Well-counter detector


Well-type ion chamber

(Radiotherapy, Brachytherapy) The well-type ion chamber, designed for brachytherapy source calibrations, is the instrument recommended for source strength measurements at the hospital level by the IAEA, see TECDOC-1274: ‘Calibration of photon and beta ray sources used in brachytherapy. Guidelines on standardised procedures at secondary standards dosimetry laboratories (SSDLs) and hospitals’. Well-type chambers, designed for brachytherapy, can in general be used both for high-dose and low-dose rate sources, as they have a large collecting volume.

The well-type chamber is easy to use, fast and reliable; the chamber is stable in response, and the measurement geometry is easy to reproduce. Special inserts are used for different types of sources; see the examples given in the following figures.

Figure W.12 shows the well-type chamber HDR 1000 from Standard Imaging with an insert for high-dose rate sources. A needle is placed in the insert, which also guides the needle to a position along the axis of the cylindrical chamber. (Notice the plastic foam to minimise heating effects.) The insert to the right is made of lead with a plastic slit in the middle, and this ‘QA tool’ can be used to determine the source position for a stepping-source afterloading unit. (Radiochromic film is also used to verify source position and step size.)

Figure W.13 shows the well-type chamber HDR 1000 Plus from Standard Imaging with an insert for low-dose rate sources. This insert is designed for stranded iodine-125 seeds (RAPID Strand, Oncura). A strand, consisting of 10 seeds in its amber coloured holder, is placed in the insert. Two measurements are made: with the strand as shown, where the upper five seeds are blocked by the cylindrical lead shield and with the strand inverted. The steel tube used for transport of the strand in its holder is standing to the left.

WFUMB
(Ultrasound) The World Federation of Ultrasound in Medicine and Biology is a federation of regional bodies covering:

- EFSUMB – Europe
- AFSUMB – Asia
- AIUM – North America
- FLAUS – Latin America
- ASUM – Australasia
- MASU – Africa

The Federation publishes a journal Ultrasound in Medicine and Biology (UMB) and hosts a website www.wfsumb.org.

White noise
(Nuclear Medicine) White noise is a random signal that in the frequency space has equal amplitude for all frequencies. The name comes from white light that contains all colours with the same intensity.

Related Article: Poisson noise

Whole-body dosimeters
(Radiation Protection) A device that allows the estimation of the whole-body effective dose incurred by the wearer.

Whole-body dosimeters are either passive (the device must be processed, and the results returned to the wearer at a later date) or active (giving an immediate reading of the dose received by the wearer).

Passive devices, such as film-badges or thermoluminescent dosimeters (TLD), are often distributed to radiographic staff and allow monitoring over regular time periods (usually 1, sometimes 2 or 3 months). The original monitors developed for whole-body monitoring were small films placed in a holder, which provided different amounts of attenuation in different regions of the film to facilitate identification of the type and energy of the radiation, which caused the exposure. TLD materials are now more commonly used for whole-body monitoring than film, although these materials are less reliable at determining the source of the radiation exposure.

More recently, one personnel dosimetry company has introduced optically stimulated materials. These have much greater sensitivity than either film or thermoluminescent (TLD) materials.

Active devices are based on semiconductor devices and may be referred to as electronic personal dosimeters. They are intended for situations where there is significant risk of exposure or where exposures are expected to be high. The original active devices were quartz-fibre electrometers. These were notoriously unreliable and have now been replaced by electronic devices using solid-state detectors. There are a variety of such solid-state devices, commonly referred to as ‘pocket dosimeters’. In addition to displaying the cumulative dose at any instant, they also have the ability to set alarms to warn the wearer when a particular exposure has been reached.

Related Articles: Personal dosimetry, Personnel dosimetry

Whole-body magnet
(Magnetic Resonance) The term whole-body magnet refers to an MRI system, which allows scanning of a whole human body, as opposed to specialised systems for brain or extremity scanning only. For whole-body MRI, the static magnetic field must be sufficiently uniform (homogeneous) in large volume allowing MRI of different body parts (e.g. brain, spine, heart, abdomen, extremities).

Whole-body radiation counters
(Radiation Protection) Whole-body radiation counting is used to measure the uptake of an administered radiopharmaceutical by patient after a suitable interval of time. It is also applied for radiation protection to estimate a total body activity of workers dealing with non-sealed radioactive sources as well as of a general public exposed to environmental radioactive contamination.

In Figure W.14, a scheme of a whole-body counter is shown. The patient is placed between a set of some scintillation counters located above and below him/her. The result of measurement presented as total example count is related to the total activity accumulated within the body. This system is called a ‘scanning shadow shield’ method.

The photograph of a whole-body counter is presented in Figure W.15.

The kind and the detailed distribution of radioisotope may be explored with using a gamma camera.

Related Articles: Contamination, Gamma camera, Radioactive source, Scintillation counter


FIGURE W.14 Scheme of a whole-body radiation counters system.

FIGURE W.15 Photograph of a whole-body radiation counters. (Courtesy of Mr. Kamil Kisielewicz, M. Sklodowska-Curie Memorial Institute, Cracow, Poland.)

Wiener spectrum
(Diagnostic Radiology) See Noise power spectrum (NPS)

Wilkinson converter
(Nuclear Medicine) See Analogue to digital converter (ADC)

Window
(Diagnostic Radiology) ‘Window’ is associated with the windowing technique used in the display of digital medical images. The need for such a technique arose initially from the observations of computed tomography (CT) images.

CT images are generally displayed on a greyscale, where more attenuating materials are represented by pixels with lighter grey shades and the less attenuating by darker ones. The standard Hounsfield scale of CT numbers ranges from approximately −1000 to +3000 HU and thus contains $2^{12}$ (4096) levels of grey (Figure W.16a). The human eye however is only capable of distinguishing about 16–100 levels of grey (depending on the person). The windowing techniques allow for the display of a range of CT numbers over the full greyscale. This way, it accommodates the limited contrast sensitivity of human eye to the high number of grey shades existing in the CT image. This technique uses two simple factors:

• Window level (WL) (or window centre) is the mid-point of the CT numbers displayed and is assigned the mean grey shade, visualising the CT number with mean grey shade (absolute grey, an analogue to optical density 1 of the characteristic film of an x-ray film)
• Window width (WW) defines the range of CT numbers around the WL, which are displayed over the greyscale

For example, if in a CT imaging system $WL = 0$ and $WW = 100$, then the water (CT = 0 H) will be displayed by medium grey, the less absorbent tissues with CT numbers from 0 to −50 will be presented with shades from medium grey to most dark grey (−50 H), and the more absorbent tissues with CT numbers from 0 to +50 will be presented with shades from medium grey to most light grey (+50 H). All tissues with densities below −50 H will be seen as black on the monitor screen, and all tissues with densities above +50 H will be seen as white. If the image is inverted (negative), the greyscale is also inverted.

The range of CT numbers displayed on the monitor can be varied by adjusting the WL and WW, according to the tissues of interest. Better differentiation between tissues (contrast) can be perceived if the window width is reduced, so that a smaller range of HU is displayed over the entire greyscale (Figure W.16b through d). In the figure,

a. The full range displayed over the greyscale
b. WL changed and WW reduced
c. WW reduced further to display a smaller range of CT numbers over the greyscale
d. WL changed to display different range of CT number values

The WL can then be set to display the tissues of interest Figure W.17.

Figure W.17 shows a typical representation of a CT scan of the lung, shown with two selections of window parameters.

The windowing technique is now used in all types of digital medical imaging systems. To obtain the best window parameters (WL and WW), the observer must first measure the pixel value (e.g. the CT number or density value in digital radiography, etc.), then adjust the WL to this pixel value and finally adjust the WW that best presents the diagnostic information. Although WW of the order of 100 will present a greyscale most suited to the human vision, many specialists prefer broad WW (say 400), as it displays many tissues and gives a better overview of the anatomical region. Once the specialist has seen the whole region with a broad window, he/she can concentrate on a specific organ with a narrower window. Some medical imaging systems can display the scan with two (or more) window parameters. For example, the image in Figure W.17a is presented with one set of window parameters (showing the lungs), but additionally, the mediastinum is selected as a region of interest, which is displayed over the image with second set of window parameters (showing only its structures).

The figures for WW and WL used in the previous examples are related to CT; however, other digital medical imaging systems may use various scales of its pixel values (e.g. −1000 to +3000 for most CT scanners and 0–512 for some digital fluoroscopy systems). This will change the numbers for WL and WW.

When an image from a digital system is photographed (say with laser film printer), the WW and WL parameters have to be shown, as otherwise the viewing conditions of this digital image

![Image](https://example.com/image.png)

**FIGURE W.16** The CT number scale in Hounsfield units (explanations in the text).
cannot be reproduced correctly at a consecutive observation of this image on the monitor.

**Related Articles:** CT number, Hounsfield scale

**Window fraction**

(Nuclear Medicine) In nuclear medicine imaging, the energy signal from each detected photon is passed through a pulse height analyser (PHA). Here, a window is set up around the photopeak energy so that photons outside that energy, which may have been scattered, are not included in the image. Most systems are set up to accept photons whose energies are within a 20% window (10% each side of the photopeak). This is known as the window fraction.

**Related Article:** Gamma camera

**Window function**

(Nuclear Medicine) A window function is used in signal processing to limit data outside a certain interval. For example, an image can be described by its frequencies and amplitudes by a Fourier transformation. Noise usually appears in the high frequency part of the Fourier spectrum. By applying a window function, all frequencies above a certain threshold are set to zero. This is commonly known as low-pass filtering. Often a window function used in this way is rolled-off by a smoothing function in order to avoid ringing artefacts.

A window function can also mimic the limited field of view of an imaging system such as a scintillation camera. This can then address aliasing effects due to overlap of periodical functions.

**Window width in single-channel analyser**

(Nuclear Medicine) In single-channel analysers, the energy window has to be selected to cover the region of the pulse height spectrum that should be analysed. The energy window is set over the photo peak corresponding to the main energy of the emitted photons.

The energy range, the window is covering, is called ‘the window width’. In a scintillation camera, the energy width should be narrow so that it excludes scattered photons to be measured.

**Wipe test**

(Radiation Protection) A wipe test is a form of monitoring for radioactive contamination, where the contamination is from a radionuclide of a type, quantity and/or energy that cannot readily be detected by hand-held contamination monitors. Wipe tests may be carried out on ‘clean’ surfaces (as part of contamination monitoring) or on sealed-sources (as part of leak-tests).

A swab sample is taken over a surface. Radioactive particles may be extracted onto the swab material. These can be detected by sensitive counting equipment. If wipe-testing for gamma- or high-beta-emitting radioisotopes, a scintillation counter is used. This device consists of a shielded well-counter, in which the sample exposes a scintillating crystal-photomultiplier tube detector.

If wipe-testing for low-energy betas or alpha emitting radioisotopes, the sample must be counted via liquid scintillation. The sample is prepared with a solvent (which absorbs the emitted radiation) and a liquid scintillant (which re-emits it as light) in a test-tube. Emitted light photons are then directly detected from the sample surface. In all cases, the samples will be counted for a time-period (from minutes to several hours) in order to allow for detectable decays from any radionuclides collected on the swab, since the activities involved may be extremely low.

**Wire cross section**

(General) The electrical properties of a copper wire are defined mainly by its cross-sectional area. Whilst mostly round in cross section, all wires with equal cross-sectional area will have similar resistance per unit length.

If diameter is $X$ mm, then cross sectional area is $1.57X^2$ mm$^2$.

Standardly available copper wires are specified by a ‘wire gauge’, of which there are many, the standard wire gauge SWG (Imperial), The American wire gauge AWG and the metric wire gauge.

Wire tables are easily found in engineering texts and on the web. However, a simple ‘rule-of-thumb’ guide to gauges, the cross-sectional areas, and the resistance per unit length can be given, based on the realisation that it is the cross-sectional area of the wire that defines its resistance per unit length:

<table>
<thead>
<tr>
<th>AWG</th>
<th>Diameter (mm)</th>
<th>Cross-sectional Area (mm$^2$)</th>
<th>Ohms per Kilometre</th>
<th>Nearest SWG</th>
<th>Nearest Metric (Whole)</th>
</tr>
</thead>
<tbody>
<tr>
<td>10</td>
<td>2.59</td>
<td>10.53</td>
<td>3.3</td>
<td>12</td>
<td>26</td>
</tr>
<tr>
<td>16</td>
<td>1.29</td>
<td>2.58</td>
<td>13.2</td>
<td>18</td>
<td>13</td>
</tr>
<tr>
<td>22</td>
<td>0.64</td>
<td>0.65</td>
<td>53</td>
<td>23</td>
<td>6</td>
</tr>
<tr>
<td>28</td>
<td>0.32</td>
<td>0.16</td>
<td>213</td>
<td>30</td>
<td>3</td>
</tr>
<tr>
<td>34</td>
<td>0.16</td>
<td>0.04</td>
<td>856</td>
<td>38</td>
<td>2</td>
</tr>
<tr>
<td>40</td>
<td>0.08</td>
<td>0.01</td>
<td>3440</td>
<td>44</td>
<td>1</td>
</tr>
</tbody>
</table>

Other handy tips in working out physical wire size to gauge number are as follows:

36 AWG has diameter exactly 0.05 inch in diameter
A change by 10 in AWG represents a 10X change in cross-sectional area
As cross-section halves, SWG number decreases by 3 approximately
As diameter halves, SWG number increases by 6 approximately
panel thin film transistor liquid crystal display in portrait mode. Usually, the workstation displays are in this mode, with its pixels – for example 5 megapixels – providing a display at least 3000 grey levels. The resolution of the monitor, which includes the diagnostic screens to observe the images during the radiological evaluation of these. These screens are optimised and subject to regular quality control. These days, the Wisconsin test cassette is used over the unattenuated scintillator to produce a film equal in density to that produced by a reference thickness of copper used over the attenuated scintillator to produce an indication of areas corresponding to various kVp (using various filters). Typical such cassette is shown in Figure W.18.

**Wisconsin test cassette**  
(Diagnostic Radiology) The Wisconsin test cassette is cassette using a single phosphor and optical attenuator. Penetration through a copper step wedge is used as an indicator of effective kVp, half-value and x-ray generator output. The degree of penetration is measured by noting the thickness of copper used over the unattenuated scintillator to produce a film equal in density to that produced by a reference thickness of copper over the attenuated scintillator. Typical such cassette is shown in Figure W.18.

**Workstation**  
(Diagnostic Radiology) The workstation is a main component of a PACS system (picture archiving and communication system), which includes the diagnostic screens to observe the images during the radiological evaluation of these. These screens are optimised and subject to regular quality control. These days, the workstations use mainly liquid crystal display (LCD) flat screens with high resolution and contrast. For example, one such screen would have contrast ratio of the order of 700:1 and display at least 3000 grey levels. The resolution of the monitor, which includes the diagnostic screens to observe the images during the radiological evaluation of these, is often presented with its pixels – for example 5 megapixels monitor (2048 × 2560). Usually, the workstation displays are in portrait mode.

**Related Articles:** Liquid crystal display, Active matrix flat panel thin film transistor liquid crystal display

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**World Health Organization (WHO)**  
(General) The World Health Organization (WHO), the directing and co-coordinating agency for health in the United Nation System, became operative in 1948. Its responsibilities include (1) providing leadership on global health matters, (2) shaping the health agenda, (3) setting norms and standards, (4) articulating evidence-based policy options, (5) providing technical support to countries in need and (6) assessing health trends.

WHO’s headquarters is based in Geneva (Switzerland) and has an organisation of six regional offices. These are located in Copenhagen (Denmark) for the European Region, in Cairo (Egypt) for the Eastern Mediterranean Region, in Brazzaville (Republic of Congo) for the African region, in New Delhi (India) for the South-East Asia region, in Manila (Philippines) for the Western Pacific region and in Washington, DC (United States) for the region of Americas. There are national offices in 147 counties. Over 8000 people including health experts, doctors, epidemiologists, scientists and administrators are employed. The budget in 2007 was about 3.3 billions USD; more than half of it coming from voluntary contribution from countries, agencies, etc.

Representatives of the member states meet every year at the World Health Assembly in Geneva to set policies and approve the budget. The work is supported by a 34-member executive board, elected at the assembly.

WHO and its member states work in collaboration with partners including other UN agencies, nongovernmental organisations and donors. WHO guidelines and standards help countries to address public health issues and tackle global health problems in order to improve people’s well-being.

Since the beginning, WHO has played an essential role in preparing the international classification of diseases and introducing the concept of ‘essential drugs’ and ‘national drug policy’. Another major activity has been the co-ordination of campaigns for the eradication of infectious diseases (e.g. the eradication of smallpox and the drastic reduction of pollio). Major efforts aim to contain malaria, tuberculosis and HIV/AIDS. Chronic diseases also receive significant support. Reducing tobacco-related deaths and diseases, and the adoption of global strategies for diet, physical activity and health rules are current objectives.

In the radiation field, WHO is a member of the Inter-Agency Committee on Radiation Safety (IACRS) and a cosponsoring organisation of the International Basic Safety Standards for Protection against Radiation and for the Safety of Radiation Sources (IAEA, 1996).

Hyperlink: http://www.who.org

**Wrap-around artefact**  
(Magnetic Resonance) Wrap-around artefact occurs when anatomy being scanned exceeds the FOV in the phase encoding direction. The anatomy outside the FOV will experience a phase shift because the phase-encoding gradient is applied across the whole body. RF signals are then detected from outside the FOV and incorrectly allocated to pixels within the FOV.

To decrease or eliminate this artefact, there are several methods that can be used. Increasing the FOV ensures that all of the anatomy is fully contained within the FOV. A smaller coil that more closely matches the FOV could be also used. Saturations bands could be applied outside the desired FOV to suppress the signal from the anatomy outside this FOV and finally the data can be oversampled, though this increases the acquisition time.

**Abbreviation:** FOV = Field of view.

**Related Article:** Signal aliasing
neuroendocrine tumours, while 125I can be used to measure blood plasma volume. Hyperpolarized 129Xe is used in magnetic resonance imaging to image ventilation in the lungs.

Xenon

(General)

Symbol Xe
Element category Non-metals
Mass number $A$ 132
Atomic number $Z$ 54
Atomic weight 131.29 g/mol
Electronic configuration $1s^2 \ 2e^2 \ 2p^6 \ 3s^2 \ 3p^6 \ 3d^{10} \ 4s^2 \ 4p^6 \ 4d^{10} \ 5s^2 \ 5p^6$
Melting point 161.4 K
Boiling point 165.1 K
Density near room temperature 5.89 kg/m$^3$

Xenon is a colourless, odourless noble gas that is mostly un-reactive. It was discovered by Sir William Ramsay and Morris Travers in 1898 and is found in trace quantities in the earth’s atmosphere (0.087 parts per million). Xenon is used, in molecular form (Xe$_2$), in excimer lasers that emit ultraviolet light at 172 and 175 nm. Such lasers are used both for eye surgery and for dermatological treatment.

Isotopes of Xenon and Their Medical Applications: Xenon has nine stable isotopes; $^{124}$Xe is used in the production of $^{123}$I and $^{125}$I, both of which are used extensively in nuclear medicine procedures. $^{123}$I is used to image the thyroid, dopamine transporters and neuroendocrine tumours, while $^{125}$I can be used to measure blood plasma volume. Hyperpolarized $^{129}$Xe is used in magnetic resonance research to image ventilation in the lungs.

Xeroradiography

(Diagnostic Radiology) Xeroradiography uses a receptor consisting of an electrically charged semiconductor plate. The image is recorded by the x-ray exposure as a pattern of varying degrees of discharge on the plate. The electrostatic image is then printed onto paper similar to the process in a Xerox copy machine (photocopier). A desirable characteristic of these images is that the process produces an edge enhancement effect and the dynamic range of the radiograph is better than those of x-ray film. The method is not used anymore, but similar effect is used in digital radiography with selenium.

X-ray

(Diagnostic Radiology) X-ray is the name given by Wilhelm Roentgen to the ‘new kind of rays’ that he discovered, investigated and demonstrated the value of for imaging the internal structures of the human body. Others suggested and promoted the name Roentgen rays.

X-rays and gamma rays occupy the same portion of the electromagnetic spectrum – approximately from $1 \times 10^{-7}$ to $1.8 \times 10^{-12}$ m wavelength.

Related Articles: Bremsstrahlung, Characteristic radiation

X-ray beam filtration

(Diagnostic Radiology) Filtration of an x-ray beam occurs whenever a beam passes through a material that has attenuation characteristics that depend on photon energies. This changes the spectrum by the selective attenuation. The common use of filtration is to remove the low-photon energy content of a beam to reduce patient exposure.

For energy spectrum filtering, materials are selected based on their attenuation characteristics in relation to photon energy. Most x-ray beams are filtered with a metal, usually aluminium, to attenuate the low photon energy end of the spectrum. The purpose is to remove the low-energy photons that generally apply radiation to the patient’s body without penetrating to contribute to image formation. K-edge filters are used, especially in mammography, to attenuate the high-energy photons for the purpose of increasing contrast sensitivity. The section of the spectrum above the K-edge energies of the filter materials is removed by the filter. Molybdenum and rhodium are the two most common filter materials used in mammography. In some applications, the spatial distribution or uniformity of an x-ray beam might be adjusted with a filter that has a varying thickness across the beam.

Related Article: Filtration total

X-ray exposure

(Diagnostic Radiology) X-ray exposure is a term used in practice to indicate either the production of an x-ray pulse (e.g. radiograph with certain duration) or radiation exposure in air – roentgen (R) or coulomb/kg.

Related Articles: Exposure, Exposure time, Exposure switch

X-ray film

(Diagnostic Radiology) One of the earliest detectors of x-rays is the x-ray film (also radiographic film). It is made of a thin clear polyester base covered with photographic emulsion. The base is about 150 μm thick, and the emulsion (silver halide crystals suspended in gelatin) is about 10 μm thick. The emulsion can be placed either on one or on both sides of the film.

Usually, the x-ray film is placed in radiographic cassette with phosphor screen (screen/film).

When the x-ray film is exposed to radiation (and light from the phosphor), the emulsion changes its chemical structure forming a latent image. After a process of film development, these changes form areas of the x-ray film, which have different opacity (or darkness) proportional to the intensity of the radiation. Following this, the film is observed on a view box.

The professional slang for x-ray (radiographic) film is simply film. For more details, see various articles describing elements of the film, its processing, characteristics, types, etc.

Related Articles: Film (as in film base, film crystals, film processing, characteristic curve, etc.)

X-ray generator

(Diagnostic Radiology) This term is mainly used to describe the combination of the high voltage generator plus the x-ray tube.

Related Articles: High voltage generator, X-ray tube
X-ray image intensifier

(Diagnostic Radiology) See Image intensifier

X-ray table exposure chamber

(Diagnostic Radiology) X-ray table exposure chamber is the ionisation chamber used in automatic exposure control (AEC) systems.

Related Article: Automatic exposure control

X-ray television

(Diagnostic Radiology) The x-ray television of a fluoroscopic system includes all parts of the imaging chain after the image intensifier. This way, the main components of the x-ray television are: the video camera tube (transforming the visual information from the output of the image intensifier into an electrical video signal); the video amplifier; the TV diagnostic monitor (or flat monitor).

This system often includes also the automatic brightness control system and/or the video gain control. As in all imaging chains, the composite (summary) modulation transfer function (MTF) of the x-ray television is the product of the MTFs of all its components:

$$\text{MTF}_{\text{sum}} = \text{MTF}_{\text{camera}} \times \text{MTF}_{\text{amplifier}} \times \text{MTF}_{\text{monitor}}$$

This way, the MTF of the worst element will define the final summary MTF. Due to this reason, care should be taken for the MTF of each element to be good. For example, the analogue TV diagnostic monitors of the x-ray television are made with high resolution. This is achieved by using double horizontal lines (raster) – 1249 lines instead of the normal 625 lines (for 50fps) or 1023 lines instead of the normal 525 lines (for 60fps). This problem does not exist in contemporary digital flat monitors (LCD or other); however, the digitizing present another problem – see article Presampling MTF.

Related Article: Presampling MTF

Further Reading: Krestel, E., ed. 1990. Imaging Systems for Medical Diagnostics, Siemens, Erlangen, Germany.

X-ray tube

(Diagnostic Radiology) The basic principle of x-rays production includes bombardment of a positively charged material (anode target) with electrons accelerated in a high voltage field (kV).

Most conventional x-ray tubes consist of a glass envelope, which contains one negative and one positive electrode – a typical ‘vacuum tube’. The negative cathode is heated in order to emit electrons (thermal electrons), which travel in the accelerating electrical field (x10kV) to the positive anode, collide with it and so produce x-ray stopping radiation (Bremsstrahlung in German). High vacuum is created in the glass envelope in order to ensure uninterrupted fly of all emitted electrons to the anode target (anode current, mA). The anode current does not depend on the high accelerating voltage (anode voltage, kV), and it depends only on the production of thermal electrons (i.e. the temperature of the cathode). Increasing the kV (anode voltage) leads to higher acceleration of the thermal electrons, which bombard the target, and thus produces x-ray quanta with higher energies (increased effective energy of the x-ray beam).

At constant kV, increasing the mA (anode current) leads to the production of higher number of x-ray quanta (increased intensity of the x-ray beam). Changing both, kV and mA, leads to the production of x-ray spectrum with different penetrating power that produces x-ray images with different contrast.

The main parts of an x-ray tube are as follows:

- Cathode – Cathode assembly that normally includes a focusing cup (Wehnelt electrode), where the heated tungsten wire (cathode filament) is placed, which produces thermal electrons. The temperature of the cathode is proportional to the flux of thermal electrons. The whole cathode assembly is under high negative potential.
- Anode – Anode assembly that includes anode stem with target plate (usually made of tungsten). There are various anode constructions. The whole anode assembly is under high positive potential. The potential between the cathode and anode (accelerating voltage) determines the energies of the created x-rays (the x-ray spectrum).
- Glass envelope (or metal envelope) that keeps high vacuum environment (minimum 10⁻⁶ mbar) in which the thermal electrons fly undisturbed from the cathode filament to the anode target.
- The anode current has to depend only on the number of thermal electrons. However, the high vacuum can extract ions from the envelope or the other components. Special care is taken for ‘degassing’ the x-ray tube, thus preventing eventual high current (arcing) between the cathode and anode that can damage the x-ray tube.

The x-ray tube is placed in an x-ray tube housing, which assures cooling of the hot anode, electrical insulation, clear path of the x-rays towards the patient and absorption of the x-rays produced in directions other than towards the patient (Figures X.1 and X.2).

Related Articles: Anode, Cathode, X-ray tube housing, Continuous spectrum, X-ray production

FIGURE X.1 Block diagram of an x-ray tube and associated high voltage (kV) transformer.

FIGURE X.2 Old-type x-ray tube with stationary anode.
X-ray tube assembly

(Diagnostic Radiology) Technical term used to describe the x-ray tube and its components.

Related Article: X-ray tube

X-ray tube housing

(Diagnostic Radiology) The x-ray tube housing provides mechanical support and protection for the tube (Figures X.3 and X.4). Most housings are made of steel or aluminium alloy. As the high voltage is supplied to tube through the cable sockets in the housing, the latest must also provide good electrical insulation. For this reason, the housing is grounded and is filled with special insulating oil (normally withstand above 220kV/cm), which provides the necessary insulation, but also assists in cooling the x-ray tube.

In stationary anode tube, the stem of the anode can be directly cooled in the oil. Often the end of the stem has attached a copper fins (radiator) for easier cooling. The low power of these tubes permits cooling through simple convection. The powerful x-ray tubes with rotating anode require active cooling. For this purpose, the heated oil is moved by an oil pump through heat exchanger (normally cooled with running water). The temperature of the circulated oil is constantly monitored by a thermal sensor, and if it exceeds certain degree (normally less than 90°C), it cuts off the tube power. Another (simpler) device is often fitted to the housing – an expansion diaphragm (rubber membrane). When the oil temperature increases, it expands and moves the membrane, which in turn activates a switch cutting-off the power until the temperature drops down (Figure X.5).

The x-ray tube generates radiation in all directions. In order to absorb this radiation, the housing is shielded inside with lead (3–4 mm). The x-ray tube housing has a small exit window covered with aluminium plate (inherent Al filtration). Through this window, the radiation is directed towards the patient (Figure X.6). However good the lead shielding is, some radiation may escape from the x-ray tube housing (leakage radiation). Measuring the leakage radiation level of the x-ray tube housing is an important safety procedure.

Related Articles: Anode, Cathode, Filament circuit, Filament current

Hyperlinks: Sprawls Foundation: http://www.sprawls.org/resources; Toshiba x-ray tube history: http://www.e-radiography.net/history/History%20of%20xray%20tubes%20toshiba.pdf
Y-90-labelled ibritumomab tiuxetan (Zevalin®)

(Nuclear Medicine) Ibritumomab tiuxetan (Zevalin®, Schering AG) is a murine IgG, kappa monoclonal antibody against the anti-CD20 antigen. This antigen is found on the cell surface of normal and malignant B lymphocytes. The antibody is produced by CHO cells and is conjugated to an MX-DTPA linker that chelates 90Y or 111In with high affinity.

90Y-labelled Zevalin is used for radioimmunotherapy of retuximab-relapsed or – refractory CD20+ follicular non-Hodgkin’s lymphoma (NHL). Pre-treatment with retuximab is performed, which leads to an advantageous biodistribution due to blocking of accessible CD20 sites in the peripheral circulation and thus elimination of B cells.

The radionuclide, Yttrium-90, is a beta emitter, of which beta particles have a maximum energy of 2.28 MeV and a maximum range of 11 mm soft tissue. The half-life is 64 h. Before treatment with 90Y-Zevalin, a pre-treatment diagnostic examination should be performed with 111Y-Zevalin, making an individual patient dose-planning possible.

The radiochemical purity should be done before injection and is assessed by instant thin-layer chromatography with silica gel (ITLC-SG) with saline as solvent. The purity should be at least 95%. The recommended infused administered activity is 15 MBq/kg body weight (maximum 1200 MBq) for patients with platelets >150 × 10^9/L and 11 MBq/kg for patients with platelet counts of 100–150 × 10^9/L.

The mean effective half-life of 90Y-Zevalin in blood is 27 h, and the urinary excretion has been reported to be only 7% during the first week. Absorbed dose to the mostly exposed organs have preliminary been reported to be spleen 2–14 mGy/Mbq, liver 2–8 mGy/Mbq, lungs 1–3 mGy/Mbq, red bone marrow 0.5–1 mGy/Mbq. The whole-body absorbed dose is estimated to 0.5 mGy/Mbq.


Yttrium (General)

<table>
<thead>
<tr>
<th>Isotope</th>
<th>90Y</th>
</tr>
</thead>
<tbody>
<tr>
<td>Half-life</td>
<td>2.67 days</td>
</tr>
<tr>
<td>Mode of decay</td>
<td>Almost pure β−-to 90Zr, some γ</td>
</tr>
<tr>
<td>Maximum decay energy, E_max</td>
<td>2.28 MeV (γ: 1.83 MeV, 0.89 MeV)</td>
</tr>
</tbody>
</table>

90Y is generator produced from its parent (90Sr) and is available with high specific activity.

The beta particles emitted from 90Y have a relatively high E_max and a maximum range in water of 11 mm, making the isotope an attractive choice for therapeutic applications.

Medical Applications: Antibody therapy – Here, 90Y is attached to monoclonal antibodies, which bind to cancer cells and deliver lethal doses of β-radiation. This application is used to treat various cancers including lymphoma, leukaemia, ovarian, colorectal, pancreatic and bone cancers.

Hepatic microsphere therapy – Hepatic microsphere therapy: a novel approach to liver cancer therapy has been developed that involves delivering synthetic microspheres loaded with 90Y directly into the liver via a catheter in placed in the hepatic artery. The microspheres are designed to both embolise the tumour’s blood supply and deliver a tumoricidal β-dose to the metastases while minimising the dose to normal liver tissue.

The relative lack of gamma emission allows treatment on an outpatient basis.

Yttrium-90 [90Y]

(Nuclear Medicine) Element: Yttrium

Isotopes: 42 < N < 60
Atomic number (Z): 39
Neutron number (N): 51
Symbol: $^{90}\text{Y}$
Production: reactor (fission)/generator

\[
^{235}\text{U}(n, f)^{90}\text{Sr} \rightarrow ^{90}\text{Y} \rightarrow ^{90}\text{Zr}
\]
or

\[
^{89}\text{Y}(n, \gamma)^{90}\text{Y} \rightarrow ^{90}\text{Zr}
\]

Daughter: $^{90}\text{Zr}$
Half-life: 2.7 days
Decay mode: $\beta^-$-decay
Radiation: $\beta^-$ 2280 keV (max) $\approx$ 760 keV (mean)
Gamma energy: none (bremsstrahlung)
Skin dose rate from 1 MBq: 100 $\mu$Sv/h at 30 cm (point source); 0.071 $\mu$Sv/h at 1 m (10 mL glass vial)
Maximum range: 11 m in air, 11 mm in water
Absorption shield: 9 mm lucite or 5 mm glass
Biological half-life: bone
Critical organ: liver, red bone marrow, bone surfaces

ALI$_{\text{min}}$ (50 mSv): 20 MBq

Effective dose: 2.7 mSv/MBq (ingestion); 1.415 mSv/MBq (inhalation)

\[2\text{D}_{3/2}
\]
\[3^9\text{Y}\]
\[88.90585\]
\[\text{[Kr]}4d5s^2\]
\[6.2173\]
\[Q_{\beta^-} 2280.1\]

Clinical Applications: Yttrium-90 is used for labelling of therapeutic radiopharmaceuticals. It is used for labelling monoclonal antibodies for radioimmunotherapy. One example is $^{89}$Y-labelled Zevalin, which is used for radioimmunotherapy of relapsed or refractory non-Hodgkin’s lymphoma (NHL).


Related Article: Y-90-labelled ibritumomab tiuxetan

Y voltage; star voltage

(90) Ye voltage; star voltage

Three-phase power is an efficient way to generate and transfer large quantities of electrical energy. Typically, the generator is mechanical in form and contains three rotating coils in a fixed magnetic field. When forced to rotate, each coil generates a sinusoidal potential, but at a phase dependent on the rotor’s physical make up. If three coils are symmetrically mounted on the rotor, each output will be 120° different in phase to the other two:

In connecting the individual coils to the supply, two different architectures are possible, the ‘Y’ formation or the star or delta form:

The Y form provides greater voltage but lower current, while the delta form provides for higher current at lower voltage.

In measuring three-phase AC voltages, it is common to measure each voltage with respect to the common earth voltage (Y-voltage). These should normally be equal, though with phases 120° apart. If they are not equal, it is usually due either a single-phase load being much greater on one phase or to some inductive external load on the supply, which causes the phases to be unbalanced). Correction usually requires engineers to rebalance the other connected to the supply.

Normal domestic single-phase supplies are taken from one of the three phases, with different buildings or areas connected to different phases to balance the load. However, when three-phase supplies are used directly, the load can put across the phases, with each part of the load receiving a larger rms voltage (1.73x) and hence power capability. Measuring between the phases should provide a higher voltage reading (for 230 V rms AC per phase, a figure of 410 V rms will be found between any two phases).

Three-phase power is usually only provided to high power loads such as motors, three-phase transformers and air-conditioning units.
Z

**Z number**

*Nuclear Medicine* See Atomic number

**Zener diode**

*Dagnostic Radiology* The Zener diode uses the Zener effect, which is a breakdown phenomenon, which holds the voltage close to a constant value called the Zener voltage.

Zener diode itself is a semiconductor with constant voltage, used as a voltage regulator, to maintain fixed voltage, in discrete components, within ICs, etc. because of its ability to maintain a constant voltage during fluctuating current conditions. It allows reverse bias current flow without damage to the avalanche region (Figure Z.1).

**Zero-crossing detector**

*Ultrasound* A simple way of detecting the dominant Doppler shift frequency is with a zero-crossing detector, whereby the number of times the oscillating demodulated Doppler signal changes sign, that is passes through zero volt (actually its mean value). The advantage of this approach is that it can be easily built; but on the other hand, it does not necessarily signify the most important parts of the frequency spectrum and it is also vulnerable to noise. The zero-crossing count is a good estimate of the mean frequency for a very narrowband signal with little noise. However, the Doppler signal from a vessel with a parabolic flow has a spectrum that is nearly rectangular distributed from zero up to the Doppler shift that corresponds to the maximum velocity. In that case, the zero-crossing detector will overestimate the mean frequency by nearly 15% and is thus a biased estimator of the mean frequency. For a more plug-like flow profile, the mean frequency approaches the RMS frequency (as detected by the zero crossing detector), and the bias is then reduced. As noise inherently increases the number of zero-crossings, it will introduce a bias. Practically, it can be estimated that the error will be at least 10% for a parabolic flow measured with a signal-to-noise ratio of 40 dB.


**Zinc cadmium sulphide**

*Dagnostic Radiology* The output screen phosphor of an image intensifier is often made from zinc cadmium sulphide (ZnS-CdS:Ag). This phosphorescence material has a broad spectrum of light emission and some afterglow, but its wavelength matches with the sensitivity of the x-ray film (in case of spot camera film). In the past, materials on this base were also used for phosphors in screen-film systems (cassettes).


**Zinc cadmium telluride**

*Dagnostic Radiology* Zinc cadmium telluride is a material used in the detector of digital mammography systems.


**Zipper artefact**

*Magnetic Resonance* The zipper artefact was one of the most common MRI equipment artefacts. It can be caused by a breakthrough of radiofrequency from an external RF source, which is then picked up by the imaging coils. It can also be caused if the receive coil picks up part of the RF excitation pulse or with the use of stimulated echoes.

The artefact will appear as an alternating dark and light line, two or three pixels in width, in the phase-encode direction. The position and width of the artefact is determined by the frequency of the interfering RF source.

---

**FIGURE Z.1** (a) Symbol and (b) typical characteristic of a 5.6 V Zener diode.
If a zipper artefact occurs, it may be due to a leak in the Faraday cage surrounding the room. If equipment such as anaesthetic monitoring and pulse oximeters are being used, they may cause a zipper artefact by allowing external RF waves to be transmitted through the Faraday cage via leads going through the waveguides into the room.

**Abbreviation:** RF = Radiofrequency

**Zonography**

*(Diagnostic Radiology)* Zonography is a radiographic tomographic procedure (linear classical tomography) in which the angle of motion of the x-ray tube is set to a low value (like 10 angular degrees) to produce relatively thick image slices. See *Linear (classical) tomography* for details.

**Related Article:** Linear (classical) tomography

**z-Sensitivity**

*(Diagnostic Radiology)* z-Sensitivity, also known as slice sensitivity profile, is a term associated with the CT slice thickness and image quality. For more detail, please see the article *Slice thickness*.

**Related Articles:** Computed tomography, Multislice scanner, Helical pitch, Helical interpolation, Partial volume effect
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