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LANDOLT- BÖRNSTEIN

Numerical Data
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GROUP VIII

VOLUME 7

Advanced Materials
and Technologies

Medical Radiological Physics

SUBVOLUME A

Fundamentals and Data in Radiobiology,
Radiation Biophysics, Dosimetry and
Medical Radiological Protection

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Numerical Data and Functional Relationships in Science and Technology
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Volume 7

Medical Radiological Physics

Subvolume A

Fundamentals and Data in Radiobiology, Radiation Biophysics,
Dosimetry and Medical Radiological Protection

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Preface

Group VIII of the New Series of Landolt-Börnstein was started a few years ago as the most recent Group of the 125 years tradition of publication numerical data and functional relationships in science and technology. It is dedicated to Advanced Materials and Technologies, and covers volumes on “Laser Physics and Applications”, “Materials”, “Energy Technologies” and “Physical Properties of Liquid Crystals”. In 2005, Volume 4 of Group VIII was edited on “Radiological Protection” against biological effects of ionizing radiations and radioisotopes. Apart from fundamental physical and biological mechanisms of ionizing radiations, Volume 4 covers sources of exposures, dosimetry, shielding against electromagnetic and in particulate ionizing radiations, and measuring techniques.

Soon after discovery of ionizing radiations, and of the phenomenon radioactivity in 1895/1896, physicians together with physicists realized the importance of these radiations both as a diagnostic tool for investigating morphological structures of organs and tissues in the human body and physiological malfunctions, as well as for treatment of uncontrolled growth of cell clusters, frequently of malignant nature such as cancer. From these early times a new medical discipline has developed in the area medical diagnosis and treatment, i.e. radiological imaging and radiooncology, as the two fundamental columns of medical radiology.

After the early decennia of medical radiology the so-called non-ionizing radiations have become growing importance in medical diagnosis and therapy. They cover ultrasound, optical radiation and lasers, electromagnetic fields as well as time-varying electric and magnetic fields. The domain of these non-ionizing radiations is in imaging, i.e. the emphasis of radiological diagnostics, although optical radiations such as lasers and ultrasonic radiation are also applied medically for therapeutic purposes.

Therefore, it was a consequent step by Springer as the Publisher of the Landolt-Börnstein New Series to publish a further volume in Group VIII dedicated to another radiological subject, namely to “Medical Radiological Physics”, as the important scientific basis of diagnostics and therapy in medical radiology.

Both the efficacy and efficiency of the above mentioned fundamental columns of medical radiology substantially depend on the kinds of radiations and energies selected for solving a specific diagnostic problem or for treatment of a tumour, as well as on the fundamental biological and biophysical effects of the various kinds of radiations and energies absorbed. Consequently, Volume VIII/7 was subdivided into two subvolumes, i.e. Subvolume 7A with a fundamental scientific discussion of kinds of radiations, biological and biophysical effects, assessment of dose to tissues, and protection of patients, clinical staff and members of the public. Subvolume 7B will be dedicated to radiological imaging, radiotherapy and treatment planning.

Subvolume VIII/7A is entitled “Fundamentals and Data in Radiobiology, Radiation Biophysics, Dosimetry and Medical Radiological Protection”, Subvolume VIII/7B will cover the tasks of medical physics in radiological diagnostics and therapy, i.e. X-ray imaging, X-ray computed tomography, nuclear medical imaging, magnetic resonance imaging, ultrasound imaging, in clinical radiotherapy, other locoregional therapies and in treatment planning.

Subvolume VIII/7A starts after an introductory chapter as guide for the reader through contents and data of the volume with four chapters, dedicated to “Radiation and Biological Effects”, “Dosimetry in Diagnostic Radiology and Radiotherapy”, “Dosimetry in Nuclear Medicine Diagnosis and Therapy”, and “Medical Radiological Protection”. Finally, radiological quantities, units and terms are explained in a Glossary, and reference is given in an Index of their location in the Chapters and relevant Sections.

Subvolume VIII/7A “Fundamentals and Data in Radiobiology, Radiation Biophysics, Dosimetry and Radiological Protection” is written by numerous internationally renowned experts, qualified in the above

scientific disciplines. Compared to most volumes in the Landolt-Börnstein Series published in the past, the present publication in the Group Advanced Materials and Technologies is not only a compilation of numerical data and functional relationships for practical purposes: As already in Volumes VIII/1 “Laser Physics and Applications” and VIII/4 “Radiological Protection”, Subvolume VIII/7A is a rather comprehensive text together with basic data, intended to be submitted to the reader on both fundamentals and corresponding data on medical radiological physics, i.e. concepts and scientific bases of medical radiology, hence radiobiology and radiation biophysics, physical dosimetry and instrumentation, medical radiological protection of patients, personnel and the general public, and medical imaging and radiotherapy.

Volume VIII/7 Subvolume A of “Medical Radiological Physics” addresses to:

- Those already working as physicians or physicists and engineers in medical radiology for getting information on most recent developments in fundamentals of this discipline, i.e. radiobiology, radiation biophysics, dosimetry, medical radiological protection and instrumentation;
- Medical physicists and engineers, participating in post-graduate education programmes in medical radiological physics, in order to become qualified experts in medical physics;
- Physicists and engineers to be qualified for future employment as health physicists or engineers in so-called competent national authorities for health protection;
- Young physicists and physicians as newcomers in the fields of dosimetry, instrumentation, radiological imaging, radiotherapy, and medical radiological protection of patients, clinical staff and members of the public.

Ultimately, the quality of the project fundamentals and data in medical physics, diagnostic radiology and radiooncology, Subvolume A of Volume VIII of the Landolt-Börnstein New Series on radiobiology, radiation physics and biophysics, dosimetry and radiological protection, is closely related to the quality of contributions received from the participants in the present multi-authored publication. The contributors have done an outstanding job in writing comprehensive treatises on very specialized subjects of various scientific disciplines. Particularly, thanks are due to the co-ordinators of the individual Chapters, Drs. J. Bernhardt, H.-M. Kramer, D. Noßke and J. Valentin, as well as to the Development Editor Dr. W. Finger and the Editorial Office of the Landolt-Börnstein New Series of Springer, Heidelberg, for the permanent and very active engagement in realizing the edition of the present Volume VIII, Subvolume A.

Finally, I may thank my wife Rosemarie for her valuable support in the technical preparation of the manuscripts for submission of the material to the Publisher.

Wolfenbüttel, January 2012

Alexander Kaul

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1 Introduction and Guide for the Reader

Contents and Data of Chapters 2-5

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1.1 Introduction

Diagnostics and therapy in medical radiology are based on fundamentals and data in radiobiology, radiation biophysics, dosimetry and radiological protection. Consequently, a crucial presupposition for the administration *lege artis* of ionizing and non-ionizing radiations as well as radiopharmaceuticals to patients in clinical radiology is the scientific pursuit and knowledge of the following facts: Kinds and physical characteristics of radiation, both ionizing and non-ionizing, their energies absorbed in biological tissues and their relation to possibly induced radiation effects and risks. State-of-the-art clinical radiology requires a thorough knowledge of these issues, and in addition an understanding of how these radiations are linked to dose, as well as full insight into this physical quantity. This is the basis of current concepts in clinical dosimetry for diagnostics and therapy, and in medical radiological protection of patients, clinical staff and of the general public.

These subjects are treated in detail in four Chapters in such a way that clinical physicists, engineers and physicians working or intending to work in radiological diagnostics and therapy get the necessary scientific background and data for a responsible professional activity in the field of medical radiology, i.e. in

- Radiations and Biological Effects;
- Dosimetry in Radiodiagnostics and Radiotherapy;
- Dosimetry in Nuclear Medicine Diagnosis and Therapy;
- Medical Radiological Protection.

The purpose of the present introduction into Volume VIII /7A is to summarize for the reader the most important fundamentals and data of the above subjects in order to guide him through the Chapters by giving him a brief overview of what he will come to know by detailed consideration of the Chapters. In addition the present introduction summarizes for the reader of the forthcoming Volume 7B the most important information of the biophysical, biological, dosimetric fundamentals and knowledge in radiological protection as the basis for a better understanding of the medical-physical elements of imaging and treatment planning in radiooncology as to be dealt with in the 2nd Volume of Medical Radiological Physics.

1.2 CHAPTER 2: Radiations and Biological Effects: Ionizing and Non-Ionizing Radiations

Ionizing radiation IR clearly refers to all kinds of radiations that have individual photon or particle energies sufficient to ionize atoms. From a puristic point of view an absolute statement like this is not possible to make regarding non-ionizing radiations NIR: Examples are low-energy photons in the order of

about 10 eV that have insufficient energy to release a bound electron from water or biological material contrary to solid-state matter; or power laser beams that can be focussed to produce plasma-ionized matter.

1.2.1 Ionizing Radiation: Physical Properties (SECT. 2.2.1)

After an introductory classification of ionizing radiations into those which deposit their energy in a medium either by direct or by indirect interactions this fundamental difference is roughly outlined as a prerequisite for understanding the energy loss of ionizing radiation traversing biological tissues.

Charged particles such as electrons, β -particles, protons, α -particles or heavier ions deposit their energy by direct Coulomb interactions, neutral particles such as photons (X-rays or γ -rays) and neutrons deposit their energy in a two-step process: In case of photons by release of electrons and - depending on the photon energy – additionally of positrons, followed by a second step of depositing the energy of the released electrons and positrons in the medium through direct Coulomb interaction. Annihilation radiation as a consequence of the eventual positron-electron recombination again triggers an indirect interaction chain in the medium.

Interaction of *photons* with matter takes place on the atomic level in the atomic shell – Photoelectric effect, Compton effect –, in the Coulomb field of the atomic nucleus – Pair production –, and in the atomic nucleus itself – Photonuclear reaction. For human tissue, Photoelectric effect and Compton effect are of equal probability at 25 keV incident photon energy. The ratio changes rapidly with increasing energies with about two Photoelectric events for 100 Compton interactions at 100 keV. For the treatment with high-energy photon radiation Pair production plays a certain role contributing a small but not negligible fraction to the therapeutic dose. Photonuclear reactions are of concern for high energy photon treatment because of the neutron production in the accelerator head and treatment room as well as in the patient itself ((γ,n) - or (γ,p) - reactions), i.e. they are primarily a radiation protection problem. A table summarizes the main characteristics such as energy dependence, Z-dependence or mass attenuation coefficient of photon interaction types as given above and to be considered for the medical application of ionizing radiation.

Charged particles in the context of medical use comprise electrons and $\beta^{-/+}$ -particles, protons, α -particles, heavy ions and negative π -mesons. Interaction of those particles is described by collision stopping power, radiation stopping power and scattering power. *Collision stopping power* for protons and heavier charged particles is described by the Bethe-Formula, modifications of the latter are needed to treat electrons (and positrons) sufficiently. Generally speaking the collision stopping power decreases with increasing energy for the non-relativistic energy region. When particles become relativistic the collision stopping power starts increasing slightly with increasing particle energy. *Radiation stopping power* describes the energy loss of the incident particle due to inelastic Coulomb scattering with the atomic nucleus of the absorbing material. Radiation stopping power increases with the square of the absorber atomic number and the kinetic energy of the incident particle but decreases with particle mass. As a consequence and for the particle energy ranges considered in medical applications only electrons exhibit significant radiative losses. Here the radiation yield for X-ray targets in the diagnostic radiology energy range is of the order of 1%, while in the megavolt energy range, e.g. for linear accelerators, it amounts to 10-20%. The *scattering power* of electrons is – in a first approximation – proportional to the square of the absorber atomic number and inversely proportional to the square of the electron kinetic energy. The *mean range of an electron* traversing matter of density 1 g/cm³ can be, in a very good approximation, numerically expressed as 50% of the incident electron energy, where the range is given in cm with kinetic electron energy quantified in MeV. *Production of charged particles* relies on different accelerator types with linear accelerators as the still predominant modality for the production of mono-energetic electrons with typical energies of 6 to 21 MeV. For the still overwhelming majority of external beam treatment cases the electron beam is converted to a bremsstrahlung photon beam with which the tumour is irradiated.

Finally, depth dose distributions of photons versus charged particles in medical applications are discussed, following directly from the physics of interactions as summarized above under the aspect of maximum tumour control at minimal side effects. Advantage of proton and heavier charged particle radiation compared to X-rays and electrons results from their *superior dose distributions* due to clinically more suitable *depth dose curves* and *narrow(er) lateral spreads*. With increasing mass of the charged particle this superiority increases with an additional biological advantage kicking in for carbon ions and beyond. However, charged particles still have to prove the translation of physical and biological superiority over photons into better clinical results.

Interaction modelling in radiation therapy is mentioned at the end of Sect. 2.2.1 on the basis of most recently published scientific literature. This subject is one of the focal points of Landolt-Börnstein Volume VIII /7B (in preparation) with the provisional title “Imaging in Radiooncology: Methods in Radiotherapy and Treatment Planning”.

1.2.2 Ionizing Radiation: Biological Effects (SECT. 2.3.1)

Acute exposures from ionizing radiations of high radiation doses and/or dose rates can cause severe tissue damage, exposure of man to low doses and/or dose rates can give rise to the induction of cancer and hereditary disease in future generations.

1.2.2.1 Deterministic Effects (SECT. 2.3.1.2)

In case of exposure to high radiation doses in the order of Gy damage may be seen quite soon after exposure, the severity of which is related to the extent of exposure. Those effects are due to obvious impairment of the functional capacity of cells, tissues and organs, and are referred to as *deterministic effects*.

The severity of those effects – such as acute depression in the numbers of circulating white blood cells and blood platelets, depression of stem cells, or such as cataracts of the eye lens as late effects – increases above the threshold where they start appearing, reflecting more cell loss and hence damage to tissue function. It also depends on individual variation in radiosensitivity in any exposed population.

Partial body irradiation of patients in radiotherapy has provided exposure data upon which to determine the tolerability of healthy tissues and organs to ionizing radiation. Under these exposure conditions unacceptable effects do not occur in more than a few percent of patients. In contrast to the precise exposure conditions of radiotherapy, exposure of workers to higher doses of low-LET radiations is most likely to be non-uniform and resulting from mixed radiations. The quantity tolerance dose therefore can at best be used as a cautious approximation to a threshold dose. Consequently the term ‘tolerance dose’ is no longer current terminology of ICRP in radiological protection.

1.2.2.2 Stochastic Effects (SECT. 2.3.1.3)

For biological effects of ionizing radiation such as cancer and inherited disorders the probability of their occurrence, but not their severity, depends on the radiation dose. In radiological protection terminology these effects are termed *stochastic effects*.

Development of cancer in tissues can be subdivided into the phases neoplastic initiation, neoplastic promotion, conversion and progression. A single mutational event in a critical gene in a single target cell *in vivo* can create the potential for a neoplastic development, which means at the level of DNA damage that there is no basis for assuming a dose threshold below which the risk of tumour induction by ionizing radiations would be zero. For radiation protection purposes a progressive increase in risk with increasing dose with no threshold is therefore assumed, not taking credit of repair mechanisms as an instrument of particular precaution in radiological protection.

1.2.2.3 Effective Dose ([SECT. 2.3.1.4](#))

The different radiosensitivities of the various organs and tissues in the human body with respect to cancer induction and mortality is described by so-called *tissue weighting factors*. Equally, the different radiation qualities of ionizing radiation as body-averaged values, each representing a mean value for the *relative biological effectiveness* of all body-tissues are considered by the ICRP. Both biological and biophysical properties – the characteristic organ and tissue radiosensitivity and biological effectiveness of the various kinds of ionizing radiations – were introduced into the concept of the ICRP of radiological protection by the idea of a primary limiting quantity, termed *effective dose*.

1.2.3 Non-Ionizing Radiations: Physical Properties ([SECT. 2.2.2](#))

For purposes of health protection and medical applications electromagnetic non-ionizing radiations NIR are divided into optical radiations of wavelengths between 100 nm and 10⁶ nm (ultraviolet, visible, infrared and laser radiation), radiofrequency radiation including microwaves of wavelengths between 1 mm and 3 000 m (corresponding to frequencies of 300 GHz and 100 kHz), and into time varying electric and magnetic fields of frequencies less than 100 kHz.

From a pragmatic point of view, magnetostatic and electrostatic fields are also dealt within the framework of NIR as is done with pressure waves thus as ultrasound and infrasound, i.e. at frequencies below 20 Hz and above 20 kHz of the audible frequency range.

Physical properties and biological effects of NIR are treated for the electromagnetic fields according to increasing frequencies, and for diagnostic ultrasound. Sources of exposure to static, time varying electric and magnetic fields of frequencies of less than 100 kHz and electromagnetic fields of frequencies above 100 kHz are investigations by nuclear magnetic resonance imaging MRI.

1.2.3.1 Diagnostic Ultrasound ([SECT. 2.2.2.2](#))

There is a large variety of diagnostic ultrasound investigations including cardiology, vascular disease, ophthalmology, internal medicine gynaecology and obstetrics. The predominant medical diagnostic application of ultrasound is in the frequency range 2-10 MHz, with occasional extension to 20 MHz for some eye examinations. Ultrasound irradiation falls into two categories – pulsed fields and continuous wave – the latter being used for Doppler frequency-shift identification of moving structures and for blood flow velocity measurement.

There are thermal and non-thermal interaction mechanisms. During clinical ultrasonographic examinations acoustic energy is transmitted into the body and interacts with its tissues in ways that may result in a measurable biological response (see [Sect. 2.3.2](#)). The best understood mechanism of interaction is that involving heating. Cavitation is a non-thermal mechanism of interaction that involves the formation, oscillation and occasional collapse of bubbles in a sound field. Ultrasound contrast agents can increase the likelihood of occurrence and the extent of ultrasound induced bioeffects *in vivo*.

1.2.3.2 Static and Slowly Varying Electric and Magnetic Fields ([SECT. 2.2.2.3](#))

Static and slowly varying electric and magnetic fields of magnetic flux densities in the order of mT have been reported e.g. inside electric trains and around cables in electric arc welding production. Strong fields are produced in high-energy technologies such as thermonuclear reactors, magneto-hydrodynamic systems and superconducting generators. The advent of the latter in the 1970s and 1980s facilitated the use of much larger magnetic fields also in medical diagnosis through the development of magnetic resonance imaging MRI and spectroscopy. The static magnetic field of MRI scanners is in routine clinical systems in the range of 0.2 to 3 T, functional MRI using magnetic fields of up to 10 T are

occasionally applied e.g. in medical research on human brain function. The following three classes of physical interactions of static magnetic fields with biological systems are well established on the basis of experimental data: electrodynamic interactions with ionic conduction currents, magnetomechanical effects, and effects on electronic spin states of reaction intermediates.

Forces and torques on both endogenous and exogenous metallic objects are the interaction mechanism of most concern. The induction of electric fields and currents in tissue is also of concern.

1.2.3.3 Time Varying Electric and Magnetic Fields of Frequencies Less Than 100 kHz and Above (SECTS. 2.2.2.4 and 2.2.5)

The main issue in time varying electric and magnetic fields of frequencies less than 100 kHz and above regarding exposures to clinicians and maintenance engineers are MRI scanners. It has been estimated that workers may be exposed to gradient magnetic fields up to 10 mT at an equivalent frequency of around 500 Hz.

Exposure to external time varying electric and magnetic fields induces electric fields and currents inside the body. The earliest therapeutic application of radiofrequency electromagnetic fields was in diathermia. Electromagnetic fields also have been used in inducing local hyperthermia for cancer therapy. Radiofrequency ablation is a technique that uses contact electrodes to deliver radiofrequency (100-500 kHz) voltages for a wide variety of medical therapies. The demand for increased spatial resolution and high signal-to-noise ratio from MRI scanners has prompted the use of much higher static magnetic fields (up to 10 T) which has led to the use of higher RF frequencies for MRI. This, in principle, augments not only the amount of RF power deposition inside the patient's body but also increases the EMF exposure for clinicians using MRI equipment and workers employed for servicing this equipment.

1.2.3.4 Optical Radiations Including Lasers (SECT. 2.2.2.6)

Medical applications of optical radiations including lasers are in dentistry, ultraviolet photo therapy, photo chemotherapy, photodynamic therapy and infrared hyperthermia. Although not directly involved in radiological imaging with the exception e.g. of laser radiation in position control of patients in percutaneous radiotherapy, properties and biological effects of optical radiation and lasers are described and discussed in the present volume, since lasers are medically applied in general surgery, ophthalmology, gastroenterology, dermatology and urology.

1.2.4 Non-Ionizing Radiations: Biological Effects (SECT. 2.3.2)

1.2.4.1 Diagnostic Ultrasound (SECT. 2.3.2.2)

Biological effects of ultrasound exposure are caused by thermal and non-thermal effects.

The amount of thermal effects of ultrasound depends on the properties of both the ultrasound field parameters and the biological tissue involving ultrasound absorption, thermal conduction and blood perfusion. Higher absorption values with the potential for undesirable ultrasound induced temperature are found in skin, tendon and spinal cord, the highest value is obtained in mineralised bone.

Fetal bone is of special concern since a relationship between the amount of ultrasound induced heating and gestational age correlates with bone development. Consequently the likelihood of producing a significant temperature increase in soft embryonic tissue is lower than in body tissue of the fetus. Fetal cerebral cortex can be significantly heated by pulsed ultrasound at diagnostic intensities, particularly in late gestation. The neurons of the developing fetal brain are considered to be most susceptible to the effects of ultrasound induced heating.

Some diagnostic equipment operated in pulsed Doppler mode has been shown to produce biologically-significant temperature increases in tissue, particularly when bone is present.

Evidence of inertial cavitation effects, i.e. existence of gas bodies in tissues from gaseous inclusions at high peak pressures, may disrupt chemical bonds and produce chemically reactive free radicals. Such evidence of inertial cavitation occurring in mammalian tissues is based on data from lithotripter exposures. It cannot be excluded that under certain clinical conditions such as surgical interventions, traumatic lesions with lesions of skin, lung or bowels, cavitation related effects such as cavitation nuclei and gas bubbles can be induced in human tissue.

1.2.4.2 Static and Slowly Varying Electric and Magnetic Fields (SECT. 2.3.2.3)

The most consistent responses in animal studies on the effects of static magnetic fields are seen in neurobehavioral studies. The results suggest that the movement of laboratory rodents in static magnetic fields equal or greater than 4T may be unpleasant for the animals, including aversive responses and conditioned avoidance. Endpoints investigated in human experimental studies of static field exposure have included peripheral nerve function, brain activity, neurobehavioral and cognitive function, sensory perception, cardiac function, blood pressure, heart rate, serum proteins and hormone levels, body and skin temperature. The results indicate that there are no effects of static magnetic field exposure at up to 8T on neurophysiological responses and cognitive functions in stationary volunteers. A dose-dependent induction of vertigo and nausea was found in workers, patients and volunteers during movement in static fields greater than about 2T. There are no reports of long-term or chronic adverse effects following prolonged static magnetic field exposure.

1.2.4.3 Time Varying Electric and Magnetic Fields of Frequencies Less Than 100 kHz (SECT. 2.3.2.4)

Exposure to power-frequency electric fields causes well-defined biological responses, ranging from perception to annoyance, through surface electric charge effects. These responses depend on the field strength, the ambient environment conditions and individual sensitivity.

High magnetic rapidly pulsed fields can induce electric fields in tissue that can directly stimulate single myelinated nerve fibres in a biologically plausible manner when the internal field strength is a few volts per meter. Such effects can arise during magnetic resonance imaging procedures.

The function of the retina as part of the CNS is affected by exposure to much weaker ELF magnetic fields than those that cause direct nerve stimulation. Threshold induced electric field strengths in extracellular fluid of the retina of 10 and 100 mV m⁻¹ at 20 Hz have been estimated to stimulate flickering light sensations, called magnetic phosphene.

Epidemiological evidence suggests that chronic low-intensity ELF magnetic field exposure is associated with an increased risk of childhood leukaemia, although a causal relationship has not been established.

1.2.4.4 Electromagnetic Fields of Frequencies Above 100 kHz (SECT. 2.3.2.5)

The photon energy of non-ionizing electromagnetic radiation is too small to affect chemical bonding directly. At 300 GHz, i.e. the boundary between microwave and infrared radiation, the photon energy is only about 10⁻³ eV (covalent bond disruption has an activation energy of 5 eV, and even hydrogen bond disruption has an activation energy of 10⁻¹ eV).

The mechanisms by which RF exposure heats biological tissue are well understood, and the most marked and consistent effect of RF exposure is that of heating, resulting in a number of heat related physiological and pathological responses: Studies suggest that cognitive functions can be adversely

affected by whole-body heat stress, resulting in increased levels of unsafe behaviour and reduced task performance.

Thermally significant RF exposure inducing temperature elevations of 1°C or more can impair male fertility and cause increased embryo and fetal losses and increase the incidence of fetal malformations and anomalies.

Cataract of the lens of eyes and opacities has not been observed in primates following either acute or prolonged exposures.

Concerning cancer-related effects, the recent in vitro and animal genotoxicity and carcinogenicity studies are rather consistent and indicate that such effects are unlikely at specific absorption rates SAR of up to 4 W kg⁻¹.

1.2.4.5 Optical Radiation and Lasers (SECT. 2.3.2.6)

Because of its limited penetration in biological tissues the direct hazardous effects of optical radiation are limited to the skin and the eyes.

Depending upon wavelengths, different types of injuries to eye and skin are:

- Photochemical injury to the cornea and conjunctiva of the eye (photokeratitis);
- Photochemical damage of the lens which may produce opacities (cataracts);
- Thermal injury to the retina of the eye from lasers or from very intensive xenon-arc sources;
- Blue light photochemical injury of the retina;
- Near-infrared thermal injury of the crystalline lens with potential for industrial heat cataract;
- Thermal injury (burns) of the skin and cornea (laser);
- Photosensitized injury of the skin.

The damage mechanism in the UV spectrum is normally photochemical, in the infrared region, it is thermal in nature.

1.3 CHAPTER 3: Dosimetry in Diagnostic Radiology and Radiotherapy

1.3.1 Diagnostic Radiology

Diagnostic radiology is responsible for the biggest contribution by far to the general public to the exposure to man-made ionizing radiation. Typical values for industrialised countries lie in the range from 1.5 to 2 mSv per year per caput of the population. This makes it necessary for the professions of those engaged in diagnostic radiology to develop and apply methods for avoiding unnecessary exposures. A determination of dose to the patient by measurements is presumed with the objectives

- to compare different methods of examinations in view of their dose requirements in equal diagnostic evidence,
- to verify conformity of a given procedure with internationally recommended dose reference values,
- to make an assessment of the possible radiation risk associated with a given procedure, and
- to avoid deterministic detriments in dose-intensive procedures as e.g. in interventional radiology.

The above objectives of measurement of the dose to patients from diagnostic radiological examinations fits into the basic system of radiological protection which is characterised by the principles justification, optimisation and ALARA, i.e. as low as reasonably achievable (see Chapter 5).

1.3.2 Radiation Therapy

The central task in radiation therapy is to deliver the “correct” dose to the patient. This means that the tumour should receive a sufficiently high dose as to terminate its existence, and keeping at the same time the danger of its recurrence at an acceptable level. According to a recommendation of the ICRU already in 1976 the dose to the tumour should be known with a relative standard uncertainty of not exceeding 2.5%. Simultaneously, excessive dose levels to the organs in the tumour’s neighbourhood must be avoided in order not to put their functionality at an unacceptable risk.

Consequently, Chapter 3 on Dosimetry in Diagnostic Radiology and Radiotherapy contains two major Sections:

- Measurement of dose in diagnostic radiology in order to estimate the dose to certain organs or the effective dose equivalent ([Sect. 3.3](#));
- Measurement of dose in external radiation therapy and in brachytherapy ([Sect. 3.4](#)).

1.3.3 Dose Measurement in Diagnostic Radiology and Radiation Therapy ([SECTS. 3.3 and 3.4](#))

Neither of the above dose quantities are routinely accessible through direct measurements. In diagnostic radiology an estimate of organ dose or effective dose to a patient is obtained from a dose measurement in free air, or on the body surface of the patient or a phantom. In external radiation therapy and in brachytherapy the dose is based on measurement of dose to water.

The basic dosimetric quantity for measurements in diagnostic radiology is the air kerma which can be measured conveniently by ionometric methods. The dosimetric quantity of interest in radiotherapy is the absorbed dose to water. The concept nowadays applied may be called a native absorbed dose to water approach that means the complete dosimetric chain is based on reference dosimeters calibrated in terms of absorbed dose to water, usually in the radiation field of a ^{60}Co source. Since older dosimetry protocols are based on reference dosimeters calibrated in the quantities air kerma or exposure it is the task of the clinical medical physicist to perform the transition from air kerma /exposure to absorbed dose to water.

1.3.3.1 Dosimetric Quantities in Diagnostic Radiology ([SECT. 3.3.2](#))

A central quantity in all forms of projection radiography is the incident air kerma being defined as the air kerma at a point in a plane corresponding to entrance surface of a specific object, e.g. the patient’s breast or a standard phantom, but in absence of the patient /phantom. Only the radiation incident on the object and not the backscattered radiation is included in the incident air kerma.

In computed tomography the geometry of the exposure is very different from that in normal projection radiography. Therefore a group of dose quantities has been specifically defined which fall under the general heading of the dose-length product of the air kerma integrated along linear section of the system.

Patient dose measurements in diagnostic radiology are undertaken either directly or indirectly. Direct measurements are undertaken at the time of the actual X-ray exposure, indirect measurements after the exposure has been made, by calculating, the patient dose from knowledge of the radiographic factors employed using calibration factors for the X-ray tube and the generator. Both direct and indirect measurements may be undertaken on physical uniform or anthropomorphic phantoms or on actual patients. Each type of measurement has a different type of role and function in the overall framework of scientific support to diagnostic radiology.

1.3.3.2 Dosimetric Quantities in Radiotherapy ([SECT. 3.4.2](#))

As indicated above, the central task in radiation therapy is to prescribe and deliver the “correct” dose to the patient, a demand which is not easy to achieve even today.

Contrary to diagnostic radiology the dosimetry for both external radiotherapy and brachytherapy is based on characterising the output from the radiation source prior to the exposure of the patient. On basis of this data the patient is exposed either for a predetermined time or a predetermined number of monitor units. In general on-patient dosimetry is performed in radiotherapy only for QA (quality assurance) or research purposes.

1.3.3.3 Dosimetric Equipment for Diagnostic Radiology ([SECT. 3.3.4](#))

[Sect. 3.3.4](#) is dedicated to a general view of dosimetric equipment for diagnostic radiology, i.e. ionization dosemeters, scintillation dosemeters, dose-area product meters, and dose-length product meters. The Section summarizes fundamental information in detail on physical principles, general characteristics of radiation detectors, and operational requirements of dosimeters given in Landolt-Börnstein Volume VIII/4 "Radiological Protection".

1.3.3.4 Dose Measurements in Diagnostic Radiology ([SECT. 3.3.5](#))

[Sect. 3.3.5](#) describes the methods of patient dose measurements in diagnostic radiology (direct and indirect measurements, patient or phantom measurements) for the various modalities, i.e. radiography, fluoroscopy, mammography, computed tomography, dental radiography (intra-oral radiography and panoramic tomography), and in interventional procedures. Employment of dosimeters and dose assessment in the various radiographic investigations are described for purposes of daily clinical practice.

1.3.3.5 Dosimetric Quantity and Equipment for Measurement in Radiotherapy ([SECT. 3.4.3](#))

[Sect. 3.4.3](#) is dedicated to the measurement of the dosimetric quantity in radiotherapy. Since absorbed dose is well correlated with the biological effects of ionizing radiation, and as the transport properties of water resemble quite closely those of many tissues and organs of the human body, the absorbed dose to water is the quantity of choice in radiation therapy. The most direct method of determining the absorbed dose to water is by means of calorimetry: The radiation induced temperature rise at a given point is proportional to the dose at that point, where the factor of proportionality is the specific heat capacity of water.

The operation of a water calorimeter needs temperature increases of the order of mK to be measured accurately with the consequence that the calorimetric measurement of the absorbed dose to water according to its definition is usually performed only at national metrology institutes. They have developed primary standards for the absorbed dose on the basis of calorimetry and also of ionometry.

Clinical dose measurements are performed according to international or national dosimetry protocols which call for the use of a reference ionization chamber calibrated in a reference field, usually ^{60}Co . These reference chambers are thimble type chambers (photon and high-energy electron fields) or parallel-plate chambers (electron fields, in particular those with mean energies below 10 MeV).

In a brief overview detectors and dosimeters are presented in [Sect. 3.4.3](#) which are employed in a clinical environment, and are listed together with their main fields of application:

- Air filled ionization chambers;
- Liquid ionization chambers and scintillators;
- Dose mapping by means of conventional silver halide or radiochromic films:
digital imaging by means of storage screens and read out of the latent image by a laser scanner;
- Thermoluminescence dosimetry (passive dosimeter);
- Alanine dosimetry (passive dosimeter) by analysing the irradiated probe by electron paramagnetic resonance in view of the density of free radicals which is proportional to the absorbed dose.

1.3.3.6 Dosimetry for Teletherapy (Sect. 3.4.4)

Since dose measurements play a central role as an input parameter for treatment planning in teletherapy, and measurement of the absorbed dose to water according to its definition is usually performed only at national metrological institutes, determination of the absorbed dose to water under clinical reference conditions is a task of exceptional responsibility of local clinical radiological physics. Consequently dosimetry for teletherapy by photons, electrons, special modalities, i.e. by tomotherapy, cyber knife, X-rays and by protons and heavier ions is treated in Sect. 3.4.4 in particular detail.

1.3.3.6.1 Photons and Electrons (Sect. 3.4.4.1)

As clinical reference conditions are usually different from the reference conditions at a calibration laboratory the calibration factor of the dosimeter for clinical measurement of absorbed dose to water in photon and electron beams is to be corrected by the ratio of the dosimeter's response under reference and user (clinical) conditions. The individual correction factors take care of the density of air, the polarity effect of the ionization chamber, the incomplete charge collection, the humidity of air, the displacement effect (due to the density differences between the air in an airfilled cavity chamber and the phantom material), the temperature effects other than the density of air, the energy dependence of response (taking into consideration properties of the radiation fields and deviations of the real ionization chamber from a perfect Bragg-Gray cavity), and deviations of the conditions of measurement from the individual clinical reference conditions.

For correction of the readings of the detectors employed in a clinical environment for the measurement of the absorbed dose to water in teletherapy Sect. 3.4.4 presents the model equation and individual equations for estimating the correction factors for a dosimeter calibrated in a calibration laboratory for the determination of the absorbed dose to water from photons and electrons under clinical reference conditions.

The percentage depth dose (PDD) or the radiation quality index defined as the tissue-phantom-ratio are determined experimentally and may be used in order to take care of the energy dependence of response of the detector used.

1.3.3.6.2 Non-Reference Conditions (Sect. 3.4.4.2)

Dosimetry protocols provide information on correction factors and their numerical values for reference conditions. Non-reference conditions are to denote situations where measurements are performed at depths and field sizes different from their reference values and off-axis. This includes measurements under so-called intensity modulated radiation therapy conditions, where the properties of the radiation field in terms of the spectral and angular distribution of particle fluences may be substantially different from what they are under reference conditions. These conditions may have an impact on the response of the detectors and make it difficult to establish a finite number of representative situations. Some photon radiation treatment machines like the Cyberknife and the Tomotherapy unit do not provide the possibility of establishing the conditions for determining standard beam specifiers as the radiation quality index or the percentage depth dose.

1.3.3.6.3 Protons and Heavier Ions (Sect. 3.4.4.4)

The advantages of use of proton and heavier ion beams in radiotherapy are low entrance and almost zero dose distal to the target as well as the lateral and distal dose gradients, both enabling better protection of normal tissues. Ions heavier than protons show a differential relative biological effectiveness (RBE) with depth which results in a higher therapy relevant dose in target tissues compared to surrounding normal tissues. Additionally, the dose of protons and heavier ions is relatively constant with depth for most of its

penetration, then rising rapidly just before the particles lose all their energy, and finally dropping sharply almost to zero.

For reference dosimetry of proton and heavier ion beams the ICRU has recommended that due to the large but not sufficiently understood differences frequently observed between Faraday Cup dosimetry and calorimeters or ionization chamber dosimetry FCs should not be used as the sole method for calibration of clinical ion beams. Water is recommended as the reference medium for the determination of absorbed dose and for beam quality measurement.

Water is also employed for relative dosimetry as it is easy to use in computerized scanning phantoms. Active and passive detectors for dosimetry under non-reference conditions or for relative dosimetry of proton and heavier ion beams must have the appropriate sensitivity, energy and dose rate independence, and spatial resolution for each clinical dosimetry task.

1.3.3.6.4 Uncertainties (SECT. 3.4.4.5)

According to the Guide to the Expression of Uncertainties in Measurement the result of a measurement of a physical quantity should be presented as a combination of the value, its uncertainty and its unit. Although this requirement generally applies to each kind of physical measurement it is of predominant importance in medical administration of ionizing radiations in tumour therapy. The statement of realistic margins of uncertainty forms the basis for an assessment of the doses in the target and also in normal tissues.

1.3.3.7 Dosimetry for Brachytherapy (SECT. 3.4.5)

Brachytherapy is a special kind of radiotherapy within the framework of oncology, where the radiation source – generally a radionuclide, occasionally a minaturized X-ray tube – is placed very close to, in contact of, or directly into the tissue, generally the tumour, which is to be treated.

High dose rate brachytherapy sources deliver dose rates of 0.2 Gy/min and more in 1 cm distance from the source, and are mostly applied for treatments of tumours of the vaginal apex, oesophagus, lungs, breasts, and prostate. Low dose rate sources are used for permanent or temporary implementation directly into the tumour for therapy of malignant diseases of oropharynx, breast carcinoma, sarcoma and in case of prostate cancer.

The quantity in terms of which a brachytherapy source is calibrated is either the reference air kerma or the air kerma strength. The quantity absorbed dose rate to water in water at a distance of 1 cm from the centre of the source is obtained by multiplying the reference air kerma rate or air kerma strength with the source type specific so-called dose rate constant.

Several protocols for source handling, calibration and dose calculation by the medical physicist in a hospital have been produced by national and international institutions. A widely accepted protocol today is the report of a Task Group 43 of the AAPM which addresses explicitly low dose rate sources, although the dose calculation algorithm is applied to high dose rate brachytherapy sources as well. According to this protocol the absorbed dose rate to water at a point at a certain distance from the source centre is proportional to the air kerma strength, the dose rate constant of the radionuclide concerned (which converts the reference air kerma rate to the absorbed dose rate to water at the reference point), and to a geometry function, which accounts for the differences in the dose rate distribution between an ideal point source and a real spatially extended source. For all source types of radionuclides used for brachytherapy consensus data are published and available for clinical medical physicists on the world wide web.

1.4 CHAPTER 4: Dosimetry in Nuclear Medicine Diagnosis and Therapy

Assessment of dose in nuclear medicine is based on estimation of the internal absorbed dose from *radiopharmaceuticals*, administered to patient for diagnostic or therapeutic purposes, and to healthy volunteers in clinical research. Internal dosimetry is equally necessary in radiological protection of workers professionally handling unsealed sources of radionuclides in clinics, and of members of the public in the control of foodstuff in cases of environmental contamination by radioisotopes.

In *diagnostic nuclear medicine* reasonably accurate dose estimates are needed for the average patient as the basis for the judgement by the physician whether the benefit of an examination justifies its possible risk. This also applies to the administration of radiopharmaceuticals to healthy volunteers in clinical research.

The *use of radiopharmaceuticals for therapy* requires even more patient- and tissue-specific dosimetry, specifically with regard to individual tumour dose estimates, and the assessment of dose to normal tissues, such as bone-marrow, liver, and kidneys.

Although *application of radiopharmaceuticals is contraindicated in pregnant women*, radioactive substances are occasionally administered to pregnant patients in case of high importance for the health of the mother or in case of unknown pregnancy. In these cases assessment of doses in the embryo or fetus is urgently necessary as a basis for the decision of an eventual interruption of pregnancy.

Another task of internal dosimetry is proper estimation of the dose to an infant as a basis for recommendation of *interruption of breastfeeding* in case of breastfeeding to the infant by a mother after administration of a radiopharmaceutical for reasons of nuclear medical diagnosis.

All of the above reasons for incorporation of unsealed radionuclides into the human body need reliable estimation of the internally absorbed dose to body organs and tissues. Other than in diagnostic administration of X-rays and in use of ionizing radiation in percutaneous and brachytherapy the absorbed dose to organs and tissues cannot be measured in internal dosimetry. Rather dosimetric and biokinetic models are needed to calculate conversion coefficients which link measurable dosimetric quantities as administered activities and kerma rate in air to absorbed dose in tissues and organs.

1.4.1 Dosimetric and Biokinetic Models (SECT. 4.2)

Consequently Chapter 4 is dedicated right at the beginning to the description of the mathematical concept of internal dosimetry as developed by the Medical Internal Radiation Dose Committee MIRD and the International Commission on Radiological Protection ICRP which has extended the MIRD dosimetric system to an age-dependent formalism for public and occupational intake of radionuclides. The subsequent Sections describe biokinetic models for deposition and retention of incorporated radionuclides in body regions, and their excretion pathways from the body.

The dosimetric MIRD-system for assessment of absorbed dose is characterized by the cumulated activity in a source region, i.e. the total number of disintegrations after uptake of a radionuclide in a source organ up to infinity, taking into account the biokinetic behaviour of the radionuclide, and the absorbed fraction of energy in a target region due to radiation emitted in the source region.

Apart from the extension of the MIRD-system by ICRP to age-dependent internal dosimetry the dosimetric concept is characterized by the so-called specific effective energy which is the equivalent dose in a target tissue per nuclear transformation in the source region considering the radiation weighting factor of the energy emitted. For adults (i.e. especially for workers) the integration time for the assessment of the committed equivalent dose is 50 years, for non-adults integration is until the age of 70 years, and for patients it may be infinite for all ages due to the usually short half-lives of radionuclides applied in nuclear medicine.

Anatomical models have first been developed as mathematical phantoms from simple ellipsoids, to age-dependent hermaphrodite phantoms of MIRD, consisting of principle organ phantoms in head and trunk, and most recently to female and male voxel phantoms on the basis of images of individuals, who were similar in height and weight to ICRP reference persons.

Special dosimetric models for tissues and organs are those for the walls of the alimentary and respiratory tract and for trabecular and cortical bone surfaces as the target tissues.

Biokinetic models describe deposition and retention of radionuclides in source regions as well as the excretion pathways from the body. In general they are first-order compartment models, each compartment representing a source region which may be an organ or tissue or part of them, or contents of the respiratory or alimentary tract, the gall bladder or the urinary bladder.

A generic model is given by ICRP for systemic activity. It describes excretion of activity directly from compartments representing the source organs instead of the physiologically more correct assumption that activity is transported back into blood and is excreted from there. In addition specifically elaborated so-called recycling models were developed by ICRP, for example for transfer of iodine in blood to the thyroid and subsequent transfer as organic iodine to other tissues and to blood again, and for physiological behaviour of iron within the body.

For the assessment of dose from inhalation and /or gastro-intestinal absorption of radionuclides the ICRP has recently developed two sophisticated biokinetic models, the Human Respiratory Tract Model HRTM and the Alimentary Tract Model HATM. The latter is an extension of a quite simple 4-compartment gastro-intestinal-tract model with mean transit times between the stomach, small intestine, upper large intestine and lower large intestine, and with absorption from the small intestine.

In contrast to the old gastro-intestinal-tract model the HATM also includes the oral cavity and the oesophagus which is important for the calculation of the effective dose. A further characteristic feature of the new model is that the absorption process of a radioactive substance is not necessarily instantaneous as it was in the old model, and that it considers retention of material in the various segments of the GI-tract with subsequent recycling of parts of the material back into the contents.

The HRTM considers several target tissues within the extrathoracic and thoracic part of the respiratory tract, i.e. the anterior nasal and posterior nasal passage, the pharynx and larynx for the extrathoracic region, and the bronchi, bronchioles as well as the alveolar interstitial region for the thoracic region of the respiratory tract.

For those who intend to devote themselves to the assessment of dose from radiopharmaceuticals in nuclear medicine diagnosis the reader is referred to a most recent published *computer programme* which can calculate doses for more than 800 radionuclides for adult females and males including pregnant females up to 9 months after conception, and for children of ages from birth to 15 years of life.

1.4.2 Biokinetic Data and Physical Properties of Radionuclides (SECTS. 4.3 and 4.4)

In a second topic, Chapter 4 is devoted to the collection of biokinetic data and to the physical properties of radionuclides which are the essential basis for estimating the internal absorbed dose from administration of radiopharmaceuticals in patients or from incorporation of radionuclides by workers or members of the general public.

With reference to and in summary of Chapter 10 of Landolt-Börnstein Volume VIII/4 “Radiological Protection” the following methods of measuring tissue, organ or whole body activity as well as of activity excreted from the human body are described under the heading “*Collection of the biokinetic data*”:

- Measurement with external probes;
- Measurement of samples of blood or excreta;
- Whole body counting;
- Gamma camera measurements by
 - Planar imaging (conjugate counting),

- SPECT and SPECT /CT,
- PET and PET /CT.

Specifically the importance of quality assurance is emphasized in collection of biokinetic data and estimation of uncertainty of their data.

Concerning *physical properties of the most important radionuclides* from diagnostic nuclear medicine rather short-lived radionuclides are used with a suitable gamma emission. Apart from presently 6 other commonly used radionuclides for diagnostics, the by far most frequently used radionuclide is ^{99m}Tc from both a dosimetric point of view and due to collimating reasons in diagnostic imaging. For PET there are presently 7 positron emitters generally available with physical half-lives between about 10 min and 15 hrs and positron energies between about 350 and nearly 1000 keV from which ^{18}F and ^{11}C are mainly used. Most important radionuclides for nuclear medical therapy are beta-emitters with physical half-lives between about 20 hrs and 50 days and energies between about 200 and 900 keV.

1.4.3 Dose Coefficients and Diagnostic Reference Activities (SECT. 4.5)

In a third topic of Chapter 4 *dose coefficients for common nuclear medicine investigations* in mSv per MBq are given to the reader. Dose coefficients (mSv / MBq) for a large number of nuclear medicine investigations are given by the ICRP for adults and children of 1, 5, 10, and 15 years of ages. Because in general there was no relevant information available the same biokinetic data have been used for children as for adults with the exception of the bladder voiding interval. However, for dose assessment specific dosimetric data have been used based upon mathematical phantoms for children.

Since the activity to the patient in nuclear medicine diagnosis should be optimized considering dose to the patient versus image quality, reference activities – *so-called diagnostic reference levels* - have been derived and established by various national and international professional bodies and radiation protection authorities. They are listed in tabular form for the reader to be used only as recommendations, e.g. in case that his own administered activities are significantly higher than those recommended or in case that the diagnostic quality of his images obtained is not satisfactory.

Due to the fact that absorbed doses to children are usually higher than the doses to adults from the same administered activity and due to the higher radiation sensitivity of children as compared to adults, a *general rule of estimating the minimum administered activity to children* of different ages for an acceptable or minimum standard of image quality is given to the reader.

1.4.4 Dosimetry in Nuclear Medical Therapy (SECT. 4.6)

Contrary to percutaneous and brachytherapy a comparable degree of accuracy of assessment of dose to the tumour is not achievable in radionuclide therapy using unsealed radiotherapeutics. Major reasons are differences in the uptake of the test dose in tumour tissue and non-tumour tissues prior to treatment as compared to the therapeutic dose, and changes of the time-dependent biokinetic behaviour of the administered radionuclide in the time during irradiation of the tumour. Nevertheless optimal treatment with unsealed radiotherapeutics needs an individual dose calculation in each patient based on the results of an investigation performed in advance to determine the individual uptake and biokinetics of the radiopharmaceutical.

This procedure is described in a 4th topic of Chapter 4 in quite detail for the nuclear physician and medical physicist in calculating the necessary activity to be incorporated into a patient for treatment of thyreotoxicosis for a given amount of absorbed dose to be administered to the thyroid, and the individual thyroid mass / volume.

Further examples for treatment and estimation of dose by nuclear medical administration of radiopharmaceuticals are given for treatment of thyroid cancer with ^{131}I -iodide, therapy of Polycytemia

vera with ^{32}P , irradiation of skeleton metastases from prostate and breast tumours for palliative therapy of bone pain, and by use of tumour-specific monoclonal antibodies labelled with radioisotopes for a possible eradication of micro metastases and disseminated tumours, and the treatment of neuro-endocrine and liver tumours.

1.4.5 Necessity of Patient-Specific Dose Planning in Radionuclide Therapy (SECT. 4.7)

As in percutaneous radiation therapy the basic goal of radionuclide therapy is to deliver sufficient radiation dose to the malignant cells in the target tissue in order to destroy these cells without causing undesired effects in normal tissue surrounding the tissue to be treated. However, usually this aim is not always attainable, and, if at all, only in quite rare cases and under presupposition of patient-individualised dosimetry.

Major reasons for individual dosimetry are:

- A pronounced variability in tumour and normal tissue uptake and biokinetics of a radiopharmaceutical;
- Patient-individual fractional distribution of a radiopharmaceutical in tumour and normal tissue.

Additionally, it is also important to note that the individual patients' organ masses may differ significantly from those of the "reference patient" in standard dosimetry models. Consequently, for dosimetry in radiopharmaceutical therapy organ and tumour masses have to be estimated for individual dose planning purposes whenever possible.

1.4.6 Dose to the Embryo and Fetus During Pregnancy (SECT. 4.8)

In vivo nuclear medical investigations are normally contraindicated during pregnancy. The reason for that is that the absorbed dose to the embryo or fetus in later stages of pregnancy in radionuclide diagnosis is often larger compared to X-ray diagnostics or even computed tomography. This is due to the exposure from excretion of most of the administered radiopharmaceuticals via the kidney and urinary bladder. However, this has become of minor importance since most diagnostic nuclear medical procedures make increasing use of short-lived radionuclides such as $^{99\text{m}}\text{Tc}$.

Apart from external irradiation of both the embryo and fetus from radioactivity in the mother's organs and tissues, exposure of the fetus is mainly due to the placental transfer of radiopharmaceuticals to the fetus and distribution of those radionuclides in fetal tissues.

Typical mean whole body fetal doses from some commonly used radiopharmaceuticals in early pregnancy and at term are given for the reader in a table with 4 different radionuclides in 10 different diagnostic procedures with activities ranging from about 1 (^{131}I , iodide, thyroid uptake) to nearly 1000 MBq ($^{99\text{m}}\text{Tc}$, DTPA, examination of renal function). In addition practical recommendations are given to decrease dose to the fetus under the aspect of radiological protection without restriction of the diagnostic evidence of an investigation (maternal hydration and frequent voiding of the mother's urinary bladder).

1.4.7 Doses to Infants From Breastfeeding (SECT. 4.9)

To limit the effective dose to a child from breastfeeding after diagnostic administration of a radiopharmaceutical to a lactating mother assessment of activity in the breast milk is performed as a basis

for possible temporary interruption in case that the effective dose to the child will exceed the value of 1 mSv. For a total of nearly 50 various radiopharmaceuticals recommendations are listed for the reader in a table where criteria for decision are:

- Interruption is not essential;
- An interruption of 12 hrs with temporary discontinuation of breastfeeding with one meal is discarded to be on the safe side;
- Interruption of breastfeeding for about 3 weeks or more.

1.5 CHAPTER 5: Medical Radiological Protection

With Röntgen's discovery of X-rays in 1895 the useful implications for diagnostic medicine were realized instantly. Becquerel's identification of radioactivity 1896 and Curie's isolation of radium in 1898 broadened the spectrum of medical uses of ionizing radiations by introduction of X-rays and radioactive substances into medical diagnosis and therapy at hospitals and clinics.

Soon after the beginning of clinical use of these new kinds of radiation radiation damage to the skin of patients and serious, even fatal consequences of radiation exposure for physicians and medical workers were observed. Consequently the need for protection against this occupational hazard became obvious, and already in 1928 first recommendations on organisation of protection of occupational exposure in terms of time, distance and shielding were published by a Commission of the International Congress of Radiology, which is now known as the International Commission on Radiological Protection, ICRP. This means that the initial concern of recommendations in medical radiological protection was for occupational exposures, and the purpose of protection was seen only as preventing deterministic detriment to medical staff. Today, the main focus of recommendations is on preventing deterministic harm for patients and minimising stochastic harm for patients, members of the medical staff and members of the general public.

1.5.1 The Concept of Medical Radiological Protection (SECT. 5.2)

Today's basic principles of radiological protection and their application to medical exposures are:

- The management of systems for justification of a radiological examination and for optimisation of radiological protection of both the patient, members of the medical staff and of the general public;
- The organisation of the protection of the patient, and
- The design of installations and structural shielding in work areas of diagnostic radiology and radiotherapy.

To guide the reader of Chapter 5 through the basis of medical radiological protection, its organisational concept and tasks in diagnostic radiology, nuclear medicine, in radiotherapy and therapeutic nuclear medicine, the fundamental principles of the general concept of radiological protection are presented at the very beginning.

This concept adapted to medical exposures is based on the first two principles of radiological protection of the ICRP, i.e. justification and optimisation of protection. *Justification* means that any decision that alters the radiation exposure of an individual should do more good than harm. *Optimisation of protection* means that the likelihood of incurring exposure, the number of people exposed, and the magnitude of their individual doses should all be kept as low as reasonably achievable, taking into account economic and societal factors (ALARA principle). The 3rd principle *application of dose limits* is

irrelevant for medical exposures of patients, since dose limits may reduce the effectiveness of the patient's diagnosis or treatment, thereby doing more harm than good. Dose limits are also irrelevant for comforters, carers and for volunteers in biomedical research. For these groups *dose and risk constraints* may be relevant.

1.5.2 The System of Applied Medical Radiological Protection ([SECT. 5.3](#))

This general concept of radiological protection developed by the ICRP and adapted to the system of medical radiological protection is the basis of applied radiological protection:

Management of the protection system in the clinic, facilities and equipment of protection, radiation sources, sources of occupational and public exposure and measures of protection, protective measures for patients in diagnostic and interventional radiology, in therapeutic radiology, nuclear medicine diagnosis and therapy. These subjects need not to be specifically introduced to the reader but have to be carefully studied by reading the relevant Sections of Chapter 5 by those whose major responsibility in a radiological clinic will be, or actually is, protection of the patient, members of the staff and of the general public.

1.5.3 Medical Radiological Protection in Magnetic Resonance Imaging and Diagnostic Ultrasound ([SECT. 5.4](#))

Since *magnetic resonance imaging MRI* has become an established diagnostic imaging modality, and *diagnostic ultrasound US* has been widely used in clinical medicine already for many years with increasing number of patients and new techniques with higher acoustic output, protection of patients – and in magnetic resonance imaging also of members of the staff as well as volunteers undergoing MR procedures – protective measures in MRI and application of diagnostic ultrasound, contraindications of the methods, protection of medical and technical personnel as well of members of the general public (if indicated) are given in detail in [Sect. 5.4](#). Considering practical protection in clinical application of MR and US the biophysical interaction mechanisms with body tissues and biological effects are described and discussed in great detail in Sections of Chapter 2.

Additionally international recommendations for protection of patients, medical staff and members of the public are given by the International Commission on Non-Ionizing Radiation Protection ICNIRP for limits of exposure to static magnetic fields in MRI, and by the World Federation for Ultrasound in Medicine and Biology WFUMB, which offer valuable information to the user in safe and effective application of MRI and US as imaging tools in clinical diagnostics.

1.6 Glossary and Index of Terms

Finally, a total of about 420 radiological quantities, units and biophysical/biomedical terms are explained in a Glossary, and reference is provided to terms in an Index of their location in the Chapters and Sections. The explanations of these quantities, units and terms in the Glossary are not always identical with definitions as given by national or international bodies and non-governmental organizations such as the International Commission on Radiological Protection ICRP, the International Commission on Radiation Units and Measurements ICRU, and the International Commission on Non-Ionizing Radiation Protection ICNIRP.

2.1 Introduction

Radiation and Biological Effects

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Diagnostics and therapy in medical radiology are based on fundamentals and data in radiobiology, radiation biophysics, dosimetry and radiological protection. Consequently, the scientific treatment and knowledge of these facts is an important presupposition for the application in clinical radiology for both, ionizing and non-ionizing radiation. The description of the physical characteristics of these radiations, the biophysical interaction mechanisms with biological tissues and the induction of possibly adverse health effects is the focal point of this Chapter 2.

Ionizing radiation refers to all kinds of radiations that have individual photon or particle energies sufficient to ionize atoms. The description of the physical characteristics of photons and particles, the interaction mechanisms with matter is the crucial point of [Sect. 2.2.1](#). Neutral particles such as photons (X-rays or γ -rays) and neutrons deposit their energy in a two-step process (cf. [Sect. 2.2.1.1](#)) Charged particles in the context of medical use comprise electrons and $\beta^{-/+}$ -particles, protons, α -particles, heavy ions and negative π -mesons. Interaction of those particles is described by collision stopping power, radiation stopping power and scattering power (cf. [Sect. 2.2.1.2](#)).

Further focal points in [Sect. 2.2.1](#) are

- Absorbed dose D and linear energy transfer LET ([Sect. 2.2.1.3](#)),
- Photons versus charged particles in medical applications ([Sect. 2.2.1.4](#)),
- Neutrons and π -mesons ([Sect. 2.2.1.5](#)), and
- Interaction modelling in radiation therapy treatment planning ([Sect. 2.2.1.5](#)).

Focal points in biological effects of ionizing radiations ([Sect. 2.3.1](#)) are discussion of

- Deterministic effects, i.e. loss of tissue or organ function following exposure to ionizing radiation ([Sect. 2.3.1.2](#)),
- Stochastic effects, i.e. cancer induction and development ([Sect. 2.3.1.3](#)),
- Risk estimation, and
- Limiting the risk in radiological protection ([Sect. 2.3.1.4](#)).

Non-ionizing radiation (NIR) is a general term for both radiations and fields that form part of the electromagnetic spectrum having insufficient radiated energy to produce ionization in the medium through which it passes. Except for a narrow band of wavelengths, called visual radiations (light), levels of NIR normality encountered in our environment are unperceived by any of the human senses until the intensity is above a threshold level and is detected as heat or some other physical or biological stimulation.

From a pragmatic point of view, magnetostatic and electrostatic fields are also dealt within the framework of NIR. The field of NIR comprises also pressure waves such as ultrasound and infrasound including airborne ultrasound and infrasound, i.e. waves on either side of the audible frequency range (20 Hz - 20 kHz).

In Sects. 2.2.2 and 2.3.2 the different types of non-ionizing radiations will be treated as follows, whereby the order of the electromagnetic field ranges is within the sequence of increasing frequency:

- Ultrasound,
- Static and slowly varying electric and magnetic fields,
- Time varying electric and magnetic field of frequencies less than 100 kHz,
- Electromagnetic fields of frequencies above 100 kHz,
- Optical radiation and lasers.

Sects. 2.2.2.2 to 2.2.2.6 will be treated – as far as possible - within a homogeneous structure:

- Physical characteristics,
- Sources,
- Technical and medical applications,
- Interaction mechanisms,
- Summary.

Scientific studies have indicated that exposures above certain threshold levels can cause not only detectable biological effects but potentially adverse health consequences. Since NIRs are ubiquitous in our natural living and working environment, and the biological studies conducted so far contains many inconsistencies, it is essential that scientific research continues to determine if health effects occur at low exposure levels. The biological effects of non-ionizing radiations will be treated in Sects. 2.3.2.2 to 2.3.2.6. The sections will be treated – as far as possible - within a homogeneous structure:

- Biological effects,
- Risk assessment,
- Recommendations for health protection and safety guidelines,
- Summary.

Since *magnetic resonance imaging MRI* has become an established diagnostic imaging modality, and *diagnostic ultrasound US* has been widely used in clinical medicine already for many years with increasing number of patients and new techniques with higher acoustic output, protection of patients – and in magnetic resonance imaging also of members of the staff as well as volunteers undergoing MR procedures – protective measures in MRI and application of diagnostic ultrasound, contraindications of the methods, protection of medical and technical personnel as well as of members of the general public (if indicated) are given in detail in Sect. 5.4.

2.2.1 Ionizing Radiation

Radiation and Biological Effects Kinds of Radiation

K.-U. KASCH

In contrast to non-ionizing radiation (see Sect. 2.2.2) ionizing radiation transports enough energy to ionize matter in a single interaction. The energy threshold for such an event is rather arbitrarily given with 12 eV. Ionizing radiation cannot be perceived by human senses but may result in severe consequences within biological structures (see Sect. 2.3.1).

Further classification leads to (a) directly and (b) indirectly ionizing radiation:

- (a) Charged particles such as electrons, protons etc. deposit their energy in the medium by direct Coulomb interactions
- (b) Neutral particles such as photons (X-rays and γ -rays described via the wave-particle duality) and neutrons deposit their energy through a two-step process:
 - o In a first step a charged particle is released in the medium, e.g. photons release electrons or positrons;
 - o In a subsequent step the released charged particle deposits energy to the medium through direct Coulomb interaction.

Both, directly and indirectly ionizing radiation are widely used as diagnostic and treatment modalities in modern medicine. The wanted effects here are inextricably linked with radiation protection problems for both, patients and personnel. Outside medicine, ionizing radiation is principally treated to be potentially detrimental to human health, be it in conjunction with artificial sources or natural background radiation.

The following sections treat the different ionizing radiation qualities in the order of their (as of today) decreasing importance in medical applications:

- Photons (Sect. 2.2.1.1),
- Charged particles (electrons, protons and heavy ions, Sects. 2.2.1.2 ff.),
- Neutron and π -mesons (Sect. 2.2.1.5).

The focus will be on the relevant physics, followed by the production and medical application of the respective radiation quality.

2.2.1.1 Photons

Photons constitute by far the most important ionizing radiation type in nature and among artificial sources as utilized in medicine and other technical applications. In physics nomenclature, photons are leptons with zero rest mass and no charge. In medical applications, their energy varies between 30 keV and 20 MeV.

Interaction of photons with matter is indirectly and can be classified on the atomic level:

- Atomic shell
 - o Coherent scattering → Rayleigh scattering, Thomson scattering
 - o Photo ionization → Photoelectric effect
 - o Incoherent scattering → Compton effect
- Coulomb field of the atomic nucleus
 - o Pair production
 - o Triplet production
- Atomic nucleus
 - o Photonuclear reaction

2.2.1.1.1 Coherent scattering

Coherent scattering of electromagnetic waves at electrical charges (Thomson scattering, cf. Fig. 2.2.1.1) or the atomic shell (Rayleigh scattering) involves no energy loss. In both processes the incident photon changes direction. Rayleigh scattering leads to an excited atomic shell that consequently emits a photon with the same energy as the incident photon. Coherent scattering presumes high atomic numbers and/or low photon energies. For human tissue, the energy threshold is about 20 keV. Therefore, coherent scattering is of limited importance in diagnostic radiology (X-rays) only.

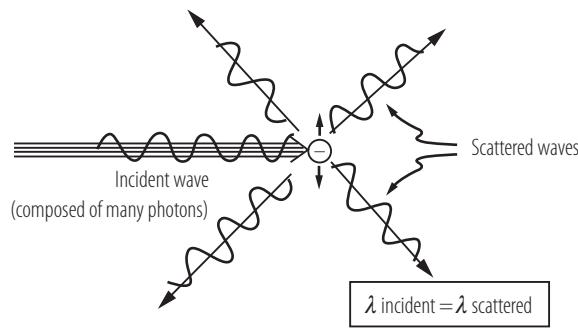


Fig. 2.2.1.1. Schematic drawing explaining Thomson scattering: Wave length (and therefore energy) are equal for incident and scattered photon.

2.2.1.1.2 Photoelectric effect

If the incident photon interacts with a tightly bound orbital electron it completely transfers its energy $E_\gamma = h\nu$ resulting to an ejected (photo) electron with a subsequent kinetic energy of $E_k = h\nu - E_B$, where $h\nu$ is the photon energy and E_B the binding energy of the electron. The probability of a photoelectric effect increases sharply at the binding energies of the attenuator or absorber material. These so called absorption edges (cf. Fig. 2.2.1.9) are used in diagnostic radiology for filtering unwanted portions of the X-ray spectrum.

The angular distribution of the photo electrons varies with the incident photon energy. While at low photon energies (up to about 100 keV) the electrons are emitted almost perpendicular to the incidence direction of the photon beam, high energy photons eject electrons almost always in forward direction.

Ejection of photo electrons results in vacancies in the (inner) atomic shells of the absorber. The repopulation of the vacancies by outer shell electrons leads to characteristic X-rays or Auger electrons, the probability for the former given by the fluorescent yield ω , while the probability for the Auger production is $1-\omega$.

The relative yield of the two competing processes depends on the atomic number Z of the absorber material and is shown in Fig. 2.2.1.2 for the K-shell. In biological tissues ($Z \sim 7$) the energy liberated due to inner shell repopulation is transferred almost completely to Auger electrons. These predominantly low-energy electrons with their limited range contribute significantly to the local energy (dose) distribution.

The atomic cross section $_{\text{a}}\tau$ for the photoelectric effect depends on the incident photon energy E_γ and atomic number Z , being proportional to $Z^4 E_\gamma^{-7/2}$. For high energies ($E_\gamma \gg m_e c^2$; i.e. the rest energy of an electron) it reads $Z^4 E_\gamma^{-1}$. As being discussed above and shown in Fig. 2.2.1.9 (note that here is $\tau \sim _{\text{a}}\tau$) the steady decrease in $_{\text{a}}\tau$ with increasing incident photon energy E_γ shows sharp discontinuities in $_{\text{a}}\tau$ when E_γ equals the binding energy for a particular electronic shell of the interaction material.

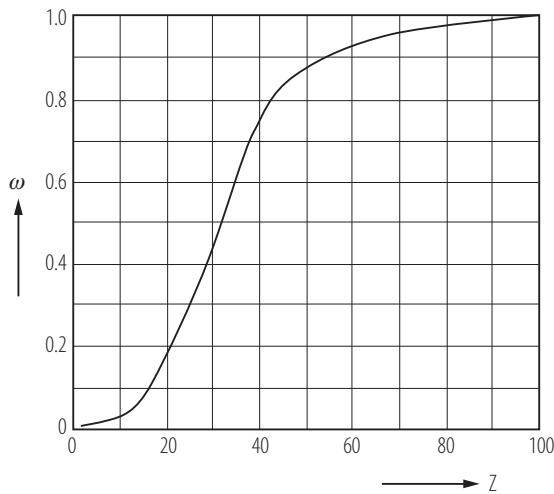


Fig. 2.2.1.2. Relative fluorescent yield ω for the K-shell as function of atomic number Z . For the yield of Auger electrons it holds: $\alpha = 1 - \omega$. For human tissue ($Z \sim 7$) the energy liberated due to inner shell repopulation is transferred almost completely to Auger electrons (modified from [05Pod]).

2.2.1.1.3 Compton effect (incoherent scattering)

While the photo-electric effect results in complete (energy) absorption of the incident photon, the Compton effect leaves a scattered photon at angle θ with energy $E'_\gamma = h\nu'$ and a recoil electron at angle ϕ with an energy E_e . Thus, the photon transfers only part of its incident energy $E_\gamma = h\nu$ (see schematic drawing in Fig. 2.2.1.3). In this process, the electron is considered to be quasi-free, i.e. the incident energy E_γ is much larger than the binding energy of the orbital electron. Considering conservation of energy and momentum in the Compton process the following relationships for the recoil electron and scattered photon can be derived [05Pod]:

$$E_e = E_\gamma \frac{\varepsilon(1 - \cos \theta)}{1 + \varepsilon(1 - \cos \theta)} \quad (2.2.1.1)$$

$$E'_\gamma = E_\gamma \frac{1}{1 + \varepsilon(1 - \cos \theta)} \quad (2.2.1.2)$$

$$\varepsilon = \frac{E_\gamma}{m_e c^2} \quad (2.2.1.3)$$

where ε is the normalized (incident) photon energy with $m_e c^2$ as rest energy of the electron.

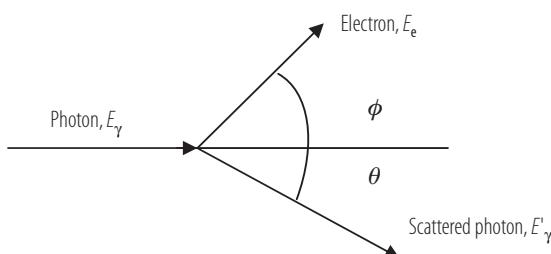


Fig. 2.2.1.3. Schematic diagram of Compton scattering. An incident photon with energy E_γ interacts with a loosely bound (essentially free) atomic electron. The electron is ejected from the atom as a recoil (Compton) electron with kinetic energy E_e and a scattered (Compton) photon with energy E'_γ is produced.

Photon scattering angle θ and electron angle ϕ are related through:

$$\cot \phi = (1 + \varepsilon) \tan(\theta/2) \quad (2.2.1.4)$$

i.e. for any photon energy the range of ϕ is between 0 and $\pi/2$ for $\theta = \pi$ (photon back scattering) and $\theta = 0$ (photon forward scattering), respectively. For a given θ it holds: the higher the incident photon energy, the smaller the recoil electron angle ϕ .

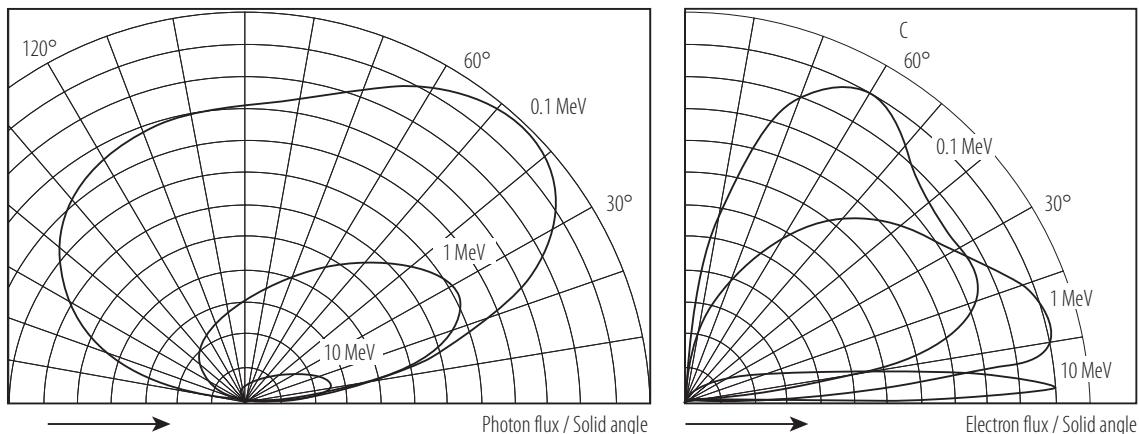


Fig. 2.2.1.4. Angular distributions of Compton photons (left, [58Eva]) and Compton electrons (right, [59Why]) normalized at the respective maximum. The arrows indicate the direction of the incident (primary) photons.

Fig. 2.2.1.4 gives a graphical representation of a more detailed analysis. Clearly visible is a "butterfly wing" for scattered photons. This is of particular interest in medical radiation protection since the Compton photons constitute the highest dose burden to operating personnel in X-ray fluoroscopy. In the energy range medical X-ray equipment is operating the produced Compton photons exhibit almost 90% of the incident photon energy (cf. [Fig. 2.2.1.5](#)).

According to [eqs. \(2.2.1.1\)](#) through to [\(2.2.1.3\)](#) the maximum energy transfer to the Compton electron occurs at $\theta = \pi$:

$$E_{e,\max} = \frac{2\epsilon E_\gamma}{1 + 2\epsilon} \quad (2.2.1.5)$$

The fraction remaining with the Compton photon is then

$$E'_{\gamma,\min} = \frac{E_\gamma}{1 + 2\epsilon} \quad (2.2.1.6)$$

Obviously, with increasing incident photon energy E_γ (i.e. increasing ϵ) the fraction given to the Compton electrons is steadily increasing. This holds in general for all angles θ and is shown in [Fig. 2.2.1.5](#) for the maximum and mean fractions of incident energy given to the recoil electrons. For example, in diagnostic radiology with mean (spectral) energies of about 30-50 keV only about 10% of the incident photon energy will be given to the Compton electron. 90% are with scattered photons. Given their angular distribution (cf. [Fig. 2.2.1.4](#)), this can constitute a significant problem of radiation protection in X-ray fluoroscopy.

A closer analysis of the Compton process employing relativistic quantum mechanics shows a preferred production of electrons with high or very low energy, while the probability of medium energy electrons is lower. This leads to the Klein-Nishina coefficients as shown in [Fig. 2.2.1.6](#).

The atomic cross section ${}^a\sigma_c$ for the Compton effect depends on the incident photon energy E_γ and atomic number Z , being proportional to $Z E_\gamma^{-1}$. Strictly speaking ${}^a\sigma_c$ breaks down into (i) the cross section ${}^a\sigma_{tr}$ of energy transfer E_{tr} to the Compton electron and (ii) the probability ${}^a\sigma_s$ for the energy E'_γ remaining with the scattered Compton photon, i.e. ${}^a\sigma_c = {}^a\sigma_{tr} + {}^a\sigma_s$ (cf. [Fig. 2.2.1.7](#)). Practically, this separation caters to the two different perspectives when a photon transverses matter: From the photon's point of view, ${}^a\sigma_c$ describes the photon attenuation in the absorber due to the Compton effect. From the absorber's perspective, on the other hand, ${}^a\sigma_{tr}$ describes the contribution of photons undergoing Compton scattering to the (local) energy deposition (see also [Sect. 2.2.1.6](#)).

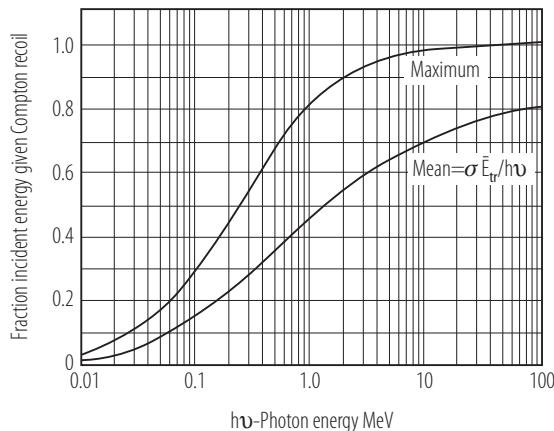


Fig. 2.2.1.5. Maximum and mean fractions of incident photon energy transferred to a Compton recoil electron in the photon energy range from 10 keV to 100 MeV (adopted from [84Joh]).

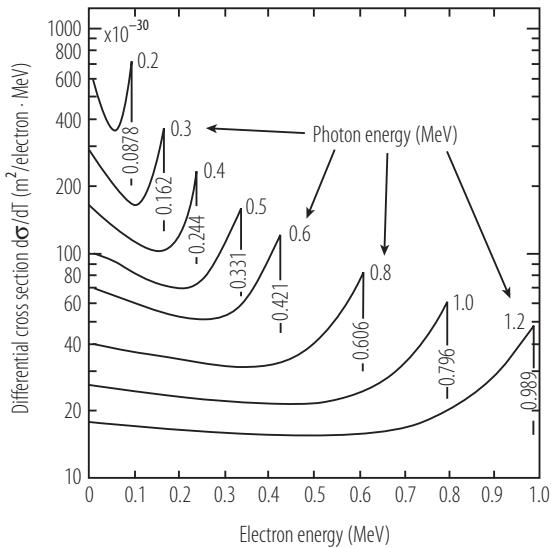


Fig. 2.2.1.6. Klein-Nishina-coefficient $d\sigma/dE_k$ as a function of kinetic energy of Compton electrons. They show a higher probability for high and very low energy electrons with lower yields for medium electron energies (adopted from ([84Joh])).

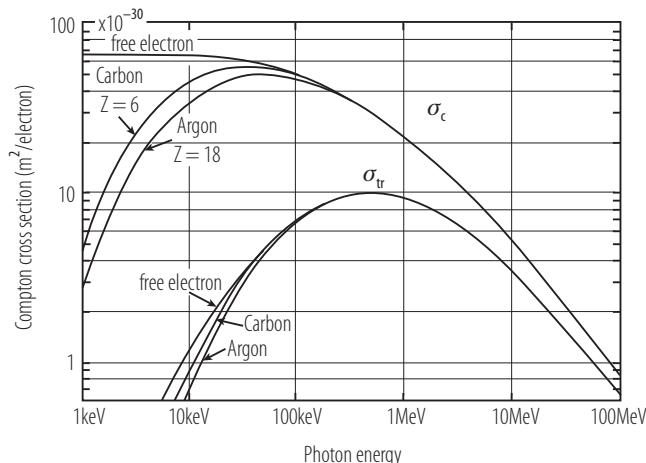


Fig. 2.2.1.7. σ_c und $a\sigma_{tr}$ (being proportional to $a\sigma_c$ and $a\sigma_{tr}$, respectively) for carbon and argon as well as for a free electron. For the latter the binding energy is neglected. The Compton effect leads to a transfer of energy, which is only in part deposited in the absorber. To describe photon attenuation, however, the total probability σ_c for Compton scattering needs to be considered. (Figure adopted from [84Joh]).

2.2.1.4 Pair production

In pair production a photon with $E_\gamma > 1.022$ MeV disappears producing an electron-positron pair with a combined kinetic energy equal to E_γ minus twice the rest energy of an electron ($m_e c^2$). When pair production occurs in the field of an orbital electron, the latter is ejected and the effect is referred to as triplet production. The available energy is shared between the electron-positron pair and the ejected electron. The threshold for triplet production is $E_\gamma > 2.044$ MeV.

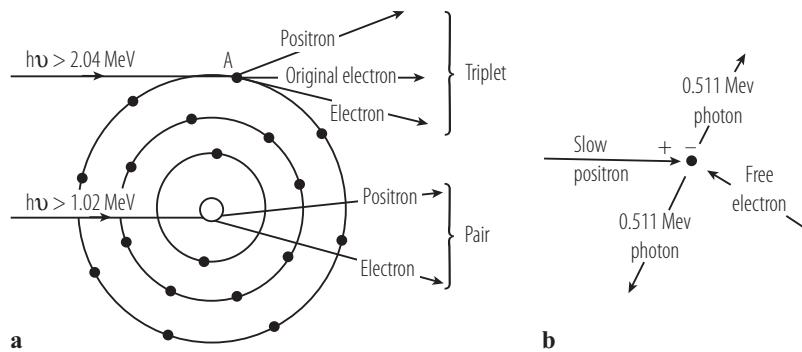


Fig. 2.2.1.8. (a) Pair and Triplet production in the Coulomb field of the nucleus and orbital electron, respectively. (b) As a consequence, the positron-electron pair annihilates producing two photons with 0.511 MeV each (adopted from [84Joh]).

For the atomic cross section of pair production it holds $\kappa \propto Z^2 \ln E_\gamma$. In the context of medical applications ($E_\gamma < 30$ MeV) pair production is only relevant in radiation protection since for human tissue ($Z \sim 7$) the probability of pair production is negligible (see also Fig. 2.2.1.11).

2.2.1.5 Photonuclear reaction

When high energy photons are absorbed by the nucleus of an atom, neutrons or protons are emitted (via (γ,n) -reactions or (γ,p) -reactions, respectively) and a transmuted (often radioactive) nuclide is produced. These photonuclear reactions are also often referred to as photodisintegration reactions. The threshold (E_γ) for a particular photonuclear reaction depends on the reaction type and the nucleus. It is in the order of 10 MeV, the binding energy of a nucleon.

Photonuclear reactions are next to negligible in photon attenuation considerations. Their probability is much smaller than that of other photon interactions. However, they are of concern in high energy photon treatment rooms because of the neutron production in the patient and the treatment room and because of the radioactivity that is induced in the treatment room air and in machine components. The neutron component should be considered in radiation oncology therapy planning for treatments with high energy photons [10Hsu]. Both neutrons and induced radioactivity pose a health hazard to personnel and must be dealt with in radiation protection. Special treatment room doors incorporating borated hydrogenous materials to thermalize and absorb the neutrons need to be used. The rooms are to be well-ventilated to reduce radioactivity in the air; treatment machine components with a low reaction cross-section and short half-life of the reaction products are to be preferred.

Table 2.2.1.1 gives an overview of possible reactions in the context of high energy photon therapy.

Table 2.2.1.1. Reaction and decay data for photonuclear reactions in high energy photon therapy.

Reaction	E_γ [MeV]	Product	Decay	$T_{1/2}$	γ -Emissions [keV]
$^{12}\text{C}(\gamma,\text{n})$	18.7	$^{11}\text{C}^*$	β^+,EC	20.4 min	511
$^{14}\text{N}(\gamma,\text{n})$	10.5	$^{13}\text{N}^*$	β^+	9.96 min	511
$^{16}\text{O}(\gamma,\text{n})$	15.6	$^{15}\text{O}^*$	β^+,EC	122 s	511
$^{16}\text{O}(\gamma,2\text{n})$	28.9	$^{14}\text{O}^*$	β^+,γ	70.6 s	511.2313

Reaction	E_γ [MeV]	Product	Decay	$T_{1/2}$	γ -Emissions [keV]
$^{27}\text{Al}(\gamma,\text{n})$	12.7	$^{26}\text{Al}^*$	$\beta^+, \text{EC}, \gamma$	6.4 s	511.1810
$^{63}\text{Cu}(\gamma,\text{n})$	10.8	$^{62}\text{Cu}^*$	β^+, EC	9.73 min	511
$^{208}\text{Pb}(\gamma,\text{n})$	7.9	^{207}Pb	stable	-	-
$^{12}\text{C}(\gamma,\text{p})$	16.0	^{11}B	stable	-	-
$^{16}\text{O}(\gamma,\text{p})$	12.1	^{15}N	stable	-	-
$^{27}\text{Al}(\gamma,\text{p})$	8.3	^{26}Mg	stable	-	-
$^{63}\text{Cu}(\gamma,\text{p})$	6.1	^{62}Ni	stable	-	-
$^{208}\text{Pb}(\gamma,\text{p})$	8.0	$^{207}\text{Tl}^*$	β^-	4.8 min	-

2.2.1.1.6 Attenuation law, energy transfer and energy deposition

For practical applications, the atomic cross section of photon interactions as described above are used to characterize the attenuation, energy transfer and energy deposition of photons in material. If ${}_a\sigma$ is the total atomic cross section, i.e.

$${}_a\sigma = {}_a\sigma_R + {}_a\tau + {}_a\sigma_c + {}_a\kappa + {}_a\sigma_{pn}$$

with

- ${}_a\sigma_R$ atomic cross section coherent scattering,
- ${}_a\tau$ atomic cross section photoelectric effect,
- ${}_a\sigma_c$ atomic cross section Compton effect,
- ${}_a\kappa$ atomic cross section pair production,
- ${}_a\sigma_{pn}$ atomic cross section photonuclear reaction,

the linear attenuation coefficient μ (in cm^{-1}) is defined as the product ${}_a\sigma \cdot n$, with n equal to the atomic volume density of the attenuator. This definition results, in analogy, in linear attenuation coefficients for any one of the various interaction types given as σ_R , τ , σ_c , κ , σ_{pn} . It should be noted that μ is strictly defined for the narrow beam approximation¹ only. Only then it holds:

$$\phi = \phi_0 e^{-\mu x} \quad (2.2.1.7)$$

with ϕ_0 and ϕ as the photon fluence proximal (in front) and distal (behind) of an attenuator of thickness x . Since $n \propto \rho$, with ρ as the physical density of the absorber material, it holds $\mu \propto \rho$. That leads to a density scaled unit, the mass attenuation coefficient: μ/ρ (in $\text{cm}^2 \text{g}^{-1}$), with

$$\phi = \phi_0 e^{-\left(\frac{\mu}{\rho}\right)\rho x} \quad (2.2.1.8)$$

with (ρx) as area density in g cm^{-2} . Fig. 2.2.1.9 shows the mass attenuation coefficient for lead with its different components.

In radiation protection, the concept of the half-value layer (HVL) and tenth-value-layer (TVL) is widely used. It describes the thickness necessary to reduce photon fluence to half or one tenth of the incident value. Using the exponential attenuation law, simple relations with the linear attenuation coefficient are found:

$$HVL = \frac{\ln 2}{\mu}, \quad TVL = \frac{\ln 10}{\mu} \quad (2.2.1.9)$$

¹ The narrow beam approximation assumes any photon undergoing an interaction to be removed from the beam. That means for instance, even if a Compton photon is scattered back into the beam it is no longer considered.

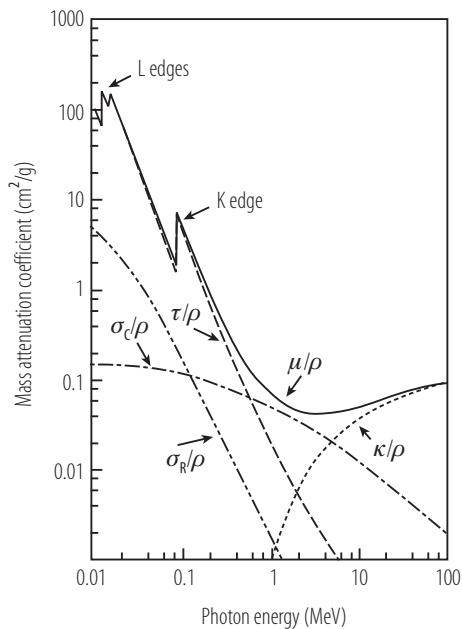


Fig. 2.2.1.9. Mass attenuation coefficient of lead as a function of incident photon energy $E_\gamma = h\nu$. Also shown are contributions of the different interaction types to the total mass attenuation coefficient. Note the absorption edges (K and L) where the atomic cross section for the photoelectric effect increases sharply (see Sect. 2.2.1.1.2). (figure adopted from [05Pod]).

While μ describes the photon attenuation by the absorber, it does not allow to quantify the energy transfer from the incident photon energy E_γ to the absorber material (e.g. the patient).

To account for energy transfer by photons to the absorber a mass energy transfer coefficient is introduced [05Kri]:

$$\frac{\mu_{\text{tr}}}{\rho} = \frac{\tau}{\rho} \left(1 - \frac{E_{\text{ch}}}{E_\gamma} \right) + \frac{\sigma_{c,\text{tr}}}{\rho} + \frac{\kappa}{\rho} \left(1 - \frac{2m_e c^2}{E_\gamma} \right) \quad (2.2.1.10)$$

with τ/ρ , $\sigma_{c,\text{tr}}/\rho$, and κ/ρ as the mass attenuation coefficients of the photoelectric effect, the energy transfer to the Compton electron (see also Sect. 2.2.1.1.3), and pair production, respectively. Photoelectric effect results in characteristic X-rays, i.e. in energy E_{ch} that is not being transferred at the point of interaction. In pair production, annihilation radiation carries energy of $2m_e c^2$ away from the interaction point.

The energy transferred by photons as described in eq. (2.2.1.10) is with secondary electrons and, in a first approximation, absorbed locally except for bremsstrahlung (radiative yield, see Sect. 2.2.1.2.3). With g as the fraction of transferred energy transformed into bremsstrahlung a mass energy absorption coefficient can be defined as:

$$\frac{\mu_{\text{en}}}{\rho} = \frac{\mu_{\text{tr}}}{\rho} (1 - g) \quad (2.2.1.11)$$

Mass attenuation coefficient μ/ρ and mass energy absorption coefficient μ_{en}/ρ represent the two different perspectives when dealing with interaction of radiation with matter: While μ/ρ describes the impact of the absorber on the traversing photon, μ_{en}/ρ indicates the impact of the photon onto the medium, i.e. the (local) energy deposition. A more detailed discussion of locally absorbed energy dose will be given in Sects. 2.2.1.3 and 2.3.1.

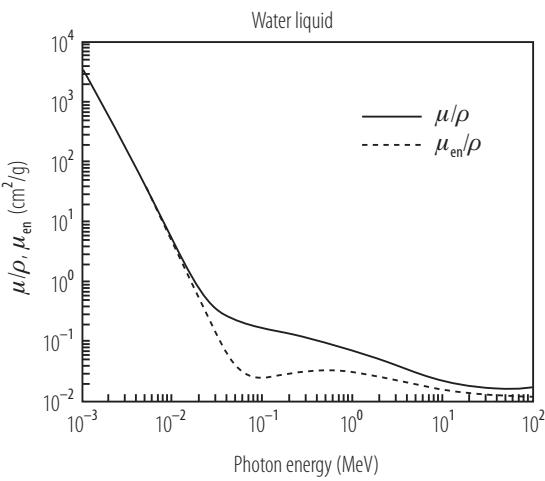


Fig. 2.2.1.10. The difference between mass attenuation coefficients and mass energy transfer coefficients in water. Only part of the energy transferred during photon attenuation is deposited locally (reprinted from [1ONIST])

2.2.1.1.7 Relative predominance of individual effects

The probability for a photon to undergo any one of the various interaction phenomena with an attenuator depends on the energy E_γ of the photon and on the atomic number Z of the attenuating material. In general, the photoelectric effect predominates at low photon energies, the Compton effect at intermediate energies and pair production at high photon energies. This is shown in Fig. 2.2.1.11 where a (E_γ , Z) diagram exhibits the two curves displaying the points for which ${}_a\tau = {}_a\sigma_c$ or ${}_a\sigma_c = {}_a\kappa$. These curves then separate the regions of relative predominance of the three most important individual effects. For example, a 60 keV photon will interact with lead ($Z = 82$) predominantly through the photoelectric effect. The same photon would interact with soft tissue ($Z_{\text{eff}} = 7.5$) predominantly through the Compton effect. These quick and simple conclusions lead to various practical consequences. For example, to amplify differences in attenuation between tissues in diagnostic radiology photon (X-ray) attenuation via photoelectric effect is favored. That needs either an energy reduction (as being done in mammography) or an increase in atomic number (i.e. using high- Z contrast media). On the other hand, when the photon energy is increasing beyond about 5 MeV, the half-value-layer HVL (see eq. (2.2.1.9) of high- Z attenuators decreases since pair production starts predominating the Compton process.

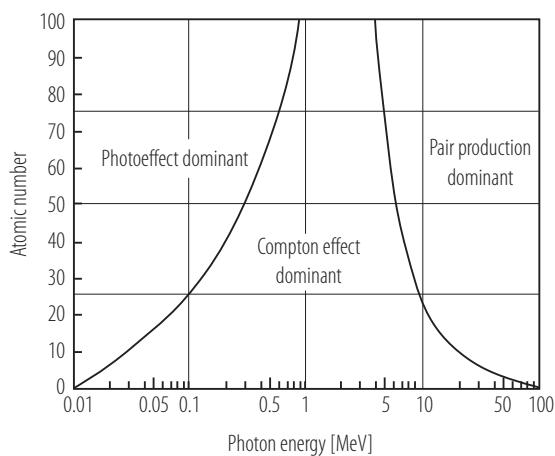


Fig. 2.2.1.11. Regional plot for the three main forms of photon interaction with matter. The left curve represents the points (energy- Z combinations) where the atomic cross sections for the photoelectric effect and the Compton effect are equal (${}_a\tau = {}_a\sigma_c$), the right curve stands for ${}_a\tau = {}_a\kappa$, i.e. for the energy- Z combinations where the atomic cross sections for the photoelectric effect and the Compton effect are equal (adopted from [05Pod]).

Table 2.2.1.2 summarizes the main photon interaction in the context of medical applications and radiation protection.

Table 2.2.1.2. Main characteristics of photon interaction types to be considered in the medical application of ionizing radiation.

	Rayleigh scattering	Photoelectric effect	Compton effect	Pair production	Photonuclear effect
Energy threshold	none	none	none	$2 m_e c^2$	$\approx 10 \text{ MeV}$
Significant energy region for tissue	< 20 keV	< 25 keV	20 keV-10 MeV	> 10 MeV	n.a.
Mode of photon interaction	Photon scattered	Photon disappears	Photon scattered	Photon disappears	Photon disappears
Particles released	None	Photoelectron	Compton electron	Electron-positron pair	Neutrons, protons
Subsequent radiation	None	Characteristic X-ray, Auger effect	Characteristic X-ray, Auger effect	Annihilation radiation	Neutrons, protons, γ
Energy dependence	E_γ^{-2}	$E_\gamma^{(-7/2...-1)}$	E_γ^{-1}	$\ln E_\gamma$	$E_\gamma > E_{\text{threshold}}$
Z-dependence atomic cross section	$a \sigma_R \propto Z^2$	$a \tau \propto Z^4$	$a \sigma_c \propto Z$	$a \kappa \propto Z^2$	Giant resonance
Z-dependence mass attenuation coeff.	$\frac{\sigma_R}{\rho} \propto Z$	$\frac{\tau}{\rho} \propto Z^3$	independent	$\frac{\kappa}{\rho} \propto Z$	Giant resonance

2.2.1.8 Photon production

Medical applications utilize two general types of photon sources, radioactive isotopes and bremsstrahlung production in X-ray tubes or linear accelerators. While radioactive isotopes offer mono-energetic photons of different energies, bremsstrahlung photons always exhibit an energy spectrum. In the majority of applications, only parts of the energy spectrum contribute to the clinical goal (imaging, cell killing, etc.). The unused parts cause side effects, i.e. cause problems in the respective context. However, there are two major advantages to the bremsstrahlungs production of photons: The energy can be tuned to the pertinent application and the equipment can be switched off with no radiation emissions left.

Radioactive isotopes (see also chapter 3)

Table 2.2.1.3 shows a selection of isotopes as used in radiation oncology. A far bigger variety is utilized in nuclear medicine. Here, very often the need arises to have pure γ -emitters for imaging purposes, i.e. to avoid concurrent particle radiation resulting in unwanted dose burden. This is achieved through so-called radio nuclide generators, where parent-daughter decay occurs in transient equilibrium. The most important generator is the $^{99}\text{Mo}-^{99m}\text{Tc}$ -type. ^{99}Mo is decaying with a branch ratio of 0.86 into ^{99m}Tc , a metastable γ -emitter with a half life of about 6 hours. A chemical reaction allows the extraction (elution) of the ^{99m}Tc ² which than, in turn, is used to label a variety of tracers for further use in nuclear medicine. After elution, it takes about 24 hours until equilibrium is re-established, a perfect time span for clinical routine operation.

² Note, that the elution of ^{99m}Tc always contains ^{99}Tc , a problem to be taken care of in the clinic but beyond the scope of this text.

Table 2.2.1.3. Physical properties of the most common isotopes used for photon production in external beam radiation therapy, brachy-therapy³, and therapeutic nuclear medicine.

Isotope	$T_{1/2}$	Decay	E_{β}^{\max} [MeV]	E_{γ} [keV]
⁶⁰ Co	5.27 a	β^-	0.331	1173, 1332
¹²⁵ I	59.2 d	EC		35, X-rays
¹³⁷ Cs	30.1 a	β^-	0.51, 1.18	662
¹⁹² Ir	73.8 d	β^-, β^-	0.24 – 0.67	296 – 612
¹⁹⁸ Au	2.69 d	β^-	0.96	410 – 680

Besides the ⁹⁹Mo-⁹⁹Tc generator, there are two more generator types of clinical importance. The ¹¹³Sn-¹¹³In-generator provides ^{113m}In, a γ -emitter with a half life of 115 days. The ¹³²Te/¹³²I-generator allows the production of ^{132m}I, a β^- -emitter with a half life of 2.3 hours.

X-ray tubes and linear accelerators (LINACs)

All diagnostic radiology is based on X-ray tubes of different specifications⁴. A typical spectrum of a clinical X-ray beam consists of a continuous bremsstrahlung spectrum (production of bremsstrahlung cf. Sect. 2.2.1.2.2) with superimposed line spectra that are characteristic of the target material as well as any attenuators placed into the beam (cf. Fig. 2.2.1.13). The bremsstrahlung spectrum originates solely in the X-ray target and is filtered so that low energy photons are suppressed. The ratio of characteristic photons to bremsstrahlung photons varies with the electron beam kinetic energy and atomic number of the target. A typical clinical X-ray tube operated at 100 keV produces one characteristic photon for every four bremsstrahlung photons. In the megavoltage range where radiation therapy machines are operated the contribution of characteristic photons to the total spectrum is negligible [05Pod].

The angular distribution of bremsstrahlung varies with electronic kinetic energy and thickness of the target. Fig. 2.2.1.12 shows typical angular distributions for different electron energies hitting a thick target. For lower energies the angular emission spectrum is becoming more and more independent of the emission direction. In a first approximation, this can also be shown for the set-up of modern X-ray tubes, i.e. for the emission spectrum in front of a thick target [84ICRU]. In the megavoltage energy range (1–50 MV) most photons are produced in forward direction.

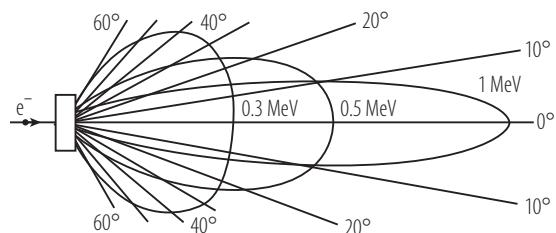


Fig. 2.2.1.12. Schematic angular distributions for different electron energies hitting a thick target (polar coordinates). For lower energies the angular emission spectrum is more and more independent of the emission direction. For high energies as being utilized in radiation therapy the emitting photons increasingly prefer the forward direction (adopted from [05Kri]).

In radiation oncology, the majority of centers uses linear accelerators (LINACs) for the production of high energy photon beams. Alternatives such as ⁶⁰Co, Betatrons or microtrons have been sidelined or even pushed into oblivion by LINACs. Using non-conservative microwave RF fields in the frequency range from 10^3 MHz to 10^4 MHz medical LINACs accelerate electrons to kinetic energies from 4 to 25 MeV. For photon production, these electrons are intercepted by a tungsten block (the target) resulting in high energy bremsstrahlung with only few superimposed characteristic X-rays, see above. Modern

³ Brachy-therapy: placing a sealed source with a radioactive isotope inside a body cavity or in tissues (see Chapter 3).

⁴ It is beyond the scope of this book to discuss frontline research such as the Carbon nanotubes used to produce X-ray spectra. However, this new CNT-technology is believed to frog leap the performance of X-ray based imaging.

medical LINACs usually provide two photon energies (e.g. 6 and 18 MV) and several electron energies (e.g. 6, 9, 12, 16, and 22 MeV).

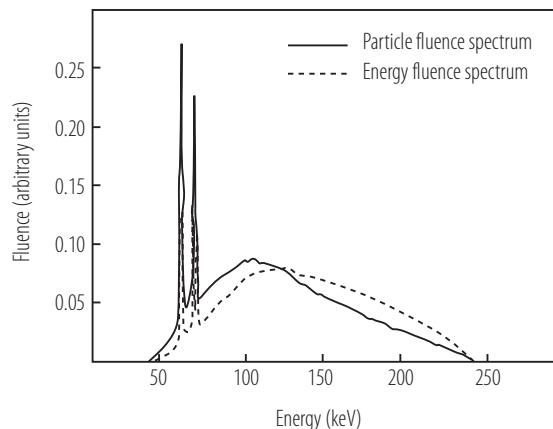


Fig. 2.2.1.13. Characteristic radiation spectra that are superimposed onto filtered bremsstrahlung spectra of a tungsten anode with X-ray tube operation at 250 kVp. Inherent filtration is 2 mm Be, added are 1 mm Al and 1.8 mm Cu filters (adopted from [05Pod]).

Table 2.2.1.4 briefly summarizes the energy ranges and production mode of photons in their three main areas of medical application.

Table 2.2.1.4. Energy ranges, production modes and radiation types as utilized in the main areas of medical applications of photons.

Energy range	Production mode	Type of Radiation	Application
20 keV – 150 keV	X-ray tubes	Bremsstrahlung, characteristic X-ray	Diagnostic radiology (imaging)
35 keV – 1 MeV	Radioactive isotopes	Mono-energetic γ or X-ray	Nuclear medicine (diagnostic & therapy), brachytherapy,
> 1 MeV	Radioactive isotope (^{60}Co), LINACs	Mono-energetic γ , bremsstrahlung radiation with charact. X-rays.	Radiation oncology

2.2.1.2 Charged particles

Charged particles in the context of medical use comprise electrons (including β -radiation), protons and heavy ions. Negative π -Mesons, a clinical hope in the 70s [77Bag] have been applied to about 1000 patients world-wide before vanishing from radiation therapy (see Sect. 2.2.1.5). Electrons and β -radiation contribute about 5% to medical applications of ionizing radiation. This fraction will increase over the next few years with Positron-Emission-Tomography (PET) Scanners becoming standard in diagnostic nuclear medicine. To a very small percentage, protons and heavy ions (e.g. ^{12}C) have been used in radiation therapy (about 80.000 patients world-wide, [10PTCOG]). However, protons and heavy ions are among the fastest growing treatment modalities today, with about thirty operational particle therapy facilities (both proton and carbon) and another twenty-three facilities or so in the planning, design, or construction stage [10PTCOG].

Typical energies for electrons (including β -radiation) in the medical context range between 2 MeV and 20 MeV, i.e. they are generally relativistic. Proton energies in therapy facilities typically range from

about 68 MeV to 250 MeV, while carbon ions may have maximum energies⁵ of 320 MeV u^{-1} to 430 MeV u^{-1} .

Charged particles are directly ionizing. Due to their charge the probability to interact is almost 100%. They experience many collisions as they traverse matter, hence their behavior is described by statistics summarizing the individual elastic and inelastic collisions with orbital electrons and nuclei. In an elastic collision the particle is deflected from its original path but no energy loss occurs, while in an inelastic collision the particle is deflected from its original path and some of its energy is transferred to an orbital electron or emitted in the form of braking radiation (bremsstrahlung). For heavy charged particles, there is a very small probability to induce nuclear reactions.

Looking at an individual particle, it interacts with matter through Coulomb interactions with atomic orbital electrons and atomic nuclei. Through these collisions the particles may lose their kinetic energy (collision and radiative losses) or change their direction of travel (scattering).

Interaction a particle undergoes with a particular atom of radius a depends on the impact parameter b of the interaction, defined as the perpendicular distance between the atomic nucleus and the particle direction before interaction (cf. Fig. 2.2.1.14).

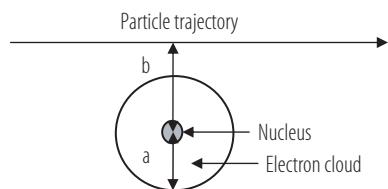


Fig. 2.2.1.14. Schematics on the interaction of a particle with an atom, where a is the atomic radius and b is the impact parameter (modified from [05Pod]).

For $b \gg a$ (cf. Fig. 2.2.1.14) the particle will undergo a soft collision with the whole atom. Elastic and inelastic scattering result in only a small amount of energy being transferred from the incident particles to the electron cloud or orbital electrons.

In Fig. 2.2.1.14 the incident particle undergoes an interaction with the Coulomb field of the atomic nucleus. If there is no energy loss, it is called Coulomb scattering or elastic nucleus scattering. In case of inelastic nucleus scattering (i.e. energy loss) the particle suffers radiative losses resulting in bremsstrahlung (braking radiation). For the energies relevant in medical applications only electrons undergo inelastic nucleus scattering.

For $b \approx \text{radius of the nucleus}$ there is a small probability of hadronic nuclear interactions. Since electrons and positrons are leptons only protons and heavy charged particles undergo occasional hadronic interaction. These may lead to fragmentation of a heavy ion traversing matter and/or hadronic cascades with subsequent secondary particles. As a consequence, the depth-dose curve (see Sect. 2.2.1.4) of heavy ion beams is characterized by a long-ranged dose tail of up to 10 - 15 % of its maximal dose, while protons preserve the desired feature of vanishing exit dose (cf. Fig. 2.2.1.17).

As for photons (see Sect. 2.2.1.1.6) interaction of charged particles with matter can either be discussed from the perspective of the absorbing matter and the impact of the traversing particle on it or, vice versa, from the particle's point of view and the impact of the absorber on its path through it. The former will be dealt with further down (cf. Sect. 2.2.1.3) the latter is quantified via stopping power and scattering power as introduced in the following sections.

2.2.1.2.1 Stopping power

2.2.1.2.1.1 Collisional losses

Collision Stopping Power describes the energy loss per unit length of the incident particle due to inelastic scattering with orbital electrons of the absorber. It is defined as:

⁵ For ions, it is customary to use the specific energy defined as the ratio of the total energy to the atomic mass number (MeV amu $^{-1}$ or MeV u $^{-1}$) [03NCRP].

$$S_{\text{coll}} = \left(\frac{dE}{dx} \right)_{\text{coll}} \text{ in } \left[\frac{\text{MeV}}{\text{cm}} \right] \quad (2.2.1.12)$$

For a non-relativistic particle with mass m and charge z the fraction of its kinetic energy E_{kin} transferred to the orbital electron can be described as:

$$\Delta E(b', E_{\text{kin}}, z, m) \propto \frac{z^2 m}{E_{\text{kin}} b'^2} \quad (2.2.1.13)$$

with b' this time defined as the perpendicular distance between orbital electron and particle direction before interaction. This relation reveals the three most important tendencies for non-relativistic particles: Energy transfer increases with the square of charge z as well as with the mass m of the particle and decreases with the kinetic energy of the incident particle.

The simple approach leading to eq. (2.2.1.13) is just the non-relativistic approximation of the more general approach embracing relativistic particles⁶, described by the Bethe-Formula [07Sie]:

$$-\left(\frac{dE}{dx} \right)_{\text{coll}} = \left(\frac{e_0^2}{4\pi\epsilon_0} \right)^2 \frac{4\pi z^2 n}{m_e v^2} \left(\ln \frac{2m_e v^2}{I(1-\beta^2)} - \beta^2 \right) \quad (2.2.1.14)$$

with:

E	kinetic energy of the particle
x	distance travelled by the particle
ϵ_0	vacuum permittivity
e_0	charge of the electron
c	speed of light
m_e	electron rest mass
v	velocity of incident particle
β	ratio v/c
z	atomic number of incident particle
n	electron density per volume unit of the absorber
I	mean ionization energy: $I \approx 7.6 Z(1 + 0.6 Z^{-2/3})$ eV

Strictly speaking, the Bethe formula holds only for protons and heavier charged particles. For electrons, the energy loss is slightly different due to their small mass and their indistinguishability, and since they suffer much larger losses by bremsstrahlung. Various corrections for eq. (2.2.1.14) are given in the literature [07Sie]. However, for the conclusions here these modifications have no impact.

Evaluation of eq. (2.2.1.14) shows a decrease of collision stopping power with increasing energy for the non-relativistic energy region. When particles become relativistic (for electrons beyond 1 MeV), however, the collision stopping power increases slightly with increasing particle energy (cf. Fig. 2.2.1.15).

With $n \sim \rho Z/A$ (ρ as physical density, A as relative atomic mass number, and Z as its atomic number of the absorber), it yields $S_{\text{coll}} \sim \rho Z/A$. That means, as long as it holds $Z/A \approx 0.5$ (i.e. constant electron density as for most human tissues), S_{coll} is independent of the absorber's atomic number. Only when the ratio Z/A decreases S_{coll} is falling (cf. Fig. 2.2.1.15).

The dependence of the Bethe formula on z^2 implies that particles with the same mass and energy but opposite charge (such as positrons) have the same stopping power and range⁷. Statistical fluctuations in the energy-loss process can also result in a r.m.s. (root mean square) spread in the actual range of individual mono-energetic particles (usually called “range straggling”).

⁶ Most of the heavy charged particles considered here are non-relativistic ($v/c \approx 0.6$ for a 250 MeV proton). However, electrons with a kinetic energy of 1 MeV read already $v/c \approx 0.94$.

⁷ However, departures from this prediction have been measured and theoretically explained by the inclusion of higher powers of z in the Bethe formula.

As a consequence of hard collisions with orbital electrons, the ejected electrons can have energies high enough to ionize again (those electrons are referred to as δ -electrons). Therefore, in dosimetry, interested in the local dose deposition, a Restricted Collision Stopping Power $S_{\text{coll},\Delta}$ is introduced. Here the index Δ gives the upper energy limit of the collisions considered in evaluations.

Due to $S_{\text{coll}} \sim \rho$ it is common to use the Mass Collision Stopping Power defined as:

$$\frac{S_{\text{coll}}}{\rho} = \frac{1}{\rho} \left(\frac{dE}{dx} \right)_{\text{coll}} \text{ in } \left[\frac{\text{MeV cm}^2}{\text{g}} \right] \quad (2.2.1.15)$$

2.2.1.2.1.2 Radiative losses

Radiation Stopping Power describes the energy loss per unit length of the incident particle due to inelastic Coulomb scattering with the atomic nucleus of the absorber. It is defined as:

$$S_{\text{rad}} = \left(\frac{dE}{dx} \right)_{\text{rad}} \text{ in } \left[\frac{\text{MeV}}{\text{cm}} \right] \quad (2.2.1.16)$$

The energy of emitted photons (referred to as bremsstrahlung or braking radiation) depends on the magnitude of the impact parameter b (cf. Fig. 2.2.1.14): the smaller the impact parameter, the higher the energy of the bremsstrahlung photon. Classical electrodynamics shows the emitted energy of an accelerated charged particle to be proportional to the square of acceleration.

That leads to:

- Energy emission $\sim M^{-2}$, with M as mass of the accelerated charge, i.e. for the energies considered here only electrons exhibit significant radiative losses,
- Energy emission $\sim Z^2$, i.e. radiative losses are increasing with increasing atomic numbers.

Relativistic approaches for electrons (> 1 MeV) yield [54Hei]:

$$-\left(\frac{dE}{dx} \right)_{\text{rad}} = \frac{e_0^6}{8\pi^2 \epsilon_0^3 m_e^2 h c^5} {}^a N Z^2 (E + m_e c^2) \left(\ln \frac{2(E + m_e c^2)}{m_e c^2} - \frac{1}{3} \right) \quad (2.2.1.17)$$

with:

E	kinetic energy of the incident electron
x	distance travelled by the particle
ϵ_0	vacuum permittivity
h	Planck's constant
e_0	charge of the electron
c	speed of light
m_e	electron rest mass
${}^a N$	atomic density per volume unit of the absorber
Z	atomic number of the absorber

As in the classical approach, S_{rad} increases with the square of the absorber atomic number Z . Furthermore, eq. (2.2.1.17) shows radiative losses to be increasing with the kinetic energy of the incident particle (see also Fig. 2.2.1.15).

A similar chain of arguments as in Sect. 2.2.1.2.1 leads to $S_{\text{rad}} \sim \rho$ and the introduction of a Mass Radiation Stopping Power:

$$\frac{S_{\text{rad}}}{\rho} = \frac{1}{\rho} \left(\frac{dE}{dx} \right)_{\text{rad}} \text{ in } \left[\frac{\text{MeV cm}^2}{\text{g}} \right]. \quad (2.2.1.18)$$

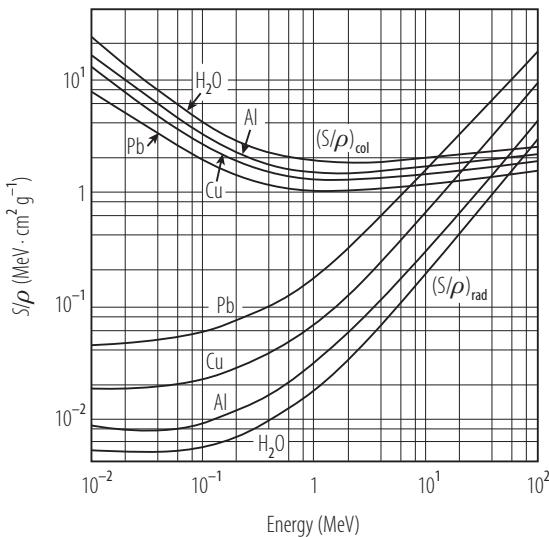


Fig. 2.2.1.15. Mass collision stopping power $(S/\rho)_{\text{col}}$ and mass radiation stopping power $(S/\rho)_{\text{rad}}$ of electrons as a function of incident (kinetic) electron energy for different materials (adopted from [05Kri]).

2.2.1.2.2 Radiative versus collisional losses

Fig. 2.2.1.15 shows both mass collision stopping power and mass radiation stopping power as functions of incident electronic energy E_{kin} and absorber material, i.e. atomic number Z . The total stopping power is given with:

$$S_{\text{tot}} = S_{\text{coll}} + S_{\text{rad}} \quad (2.2.1.19)$$

and a total mass stopping power as:

$$\frac{S_{\text{tot}}}{\rho} = \frac{S_{\text{coll}}}{\rho} + \frac{S_{\text{rad}}}{\rho} \quad (2.2.1.20)$$

In a first approximation the ratio of radiation and collision stopping power for electrons is given by:

$$\frac{\left(\frac{dE}{dx}\right)_{\text{rad}}}{\left(\frac{dE}{dx}\right)_{\text{coll}}} \approx \frac{ZE_{\text{kin}}}{1600 m_e c^2} \quad (2.2.1.21)$$

with absorber atomic number Z , kinetic electron energy E_{kin} , and $m_e c^2$ as electron rest energy. From eq. (2.2.1.21) the radiation yield for X ray targets (cf. Sect. 2.2.1.1.8) in the diagnostic radiology energy range (~100 keV) is of the order of 1%, while in the megavoltage energy range (e.g. for linear accelerators) it amounts to 10–20%.

2.2.1.2.3 Scattering of charged particles

When a charged particle penetrates an absorbing medium, most of the scattering interactions lead to small deflections originating in a large number of very small deflections. The overall result is, therefore, referred to as multiple scattering.

Multiple Coulomb scattering is treated in several publications (e.g. [93And]). However, all are based on the theory put forward by Molière as early as 1948 [48Mol]. His approach is directed at finding the angular distributions without regard to the lateral displacement. Subsequently, the original theory was slightly modified by other authors. Depending on the target materials, the best agreement with current

experimental data is being reached equally by the Bethe interpretation of the original theory [91Lyn] and the Fano modification [93Got]. As a consequence, the angular and spatial spread of a pencil beam⁸ can be approximated by a Gaussian distribution. For electrons, the International Commission on Radiation Units and Measurements (ICRU) defines a mass scattering power T/ρ as [84ICRU, 05Pod]:

$$\frac{T}{\rho} = \frac{1}{\rho} \frac{\overline{\theta^2}}{dl} \text{ or } \frac{T}{\rho} = \frac{\overline{\theta^2}}{\rho l} \quad (2.2.1.22)$$

where $\overline{\theta^2}$ is the mean square angle of scattering for electrons traversing a path length l through an absorbing medium with density ρ . $\overline{\theta^2}$ is proportional to the mass thickness ρl of the absorber.

The scattering power, in a first approximation, is proportional to the square of the absorber atomic number and inversely proportional to the square of the electron kinetic energy.

2.2.1.2.4 Range of heavy charged particles

In contrast to photons, charged particles exhibit a clearly defined, finite range. Usually, a mean range \bar{R} is given, where the transmission of particles through an absorber is decreased by 50 % (see also Fig. 2.2.1.16). Mono-energetic heavy charged particles (i.e. except electrons) exhibit only little deflections and range straggling. This gives rise to the definition:

$$\bar{R} = \int_0^{E_{\text{kin}}^{(i)}} (S_{\text{tot}}(E_{\text{kin}}))^{-1} dE_{\text{kin}} \quad (2.2.1.23)$$

where $E_{\text{kin}}^{(i)}$ is the initial kinetic energy of the particle. Since for heavy charged particles radiative losses can be neglected in the considered energy range (cf. Sect. 2.2.1.2.2) the mean range is a function of collision stopping power only, i.e.:

$$\bar{R} = \int_0^{E_{\text{kin}}^{(i)}} \left(\frac{dE}{dx}(E_{\text{kin}}) \right)_{\text{coll}}^{-1} dE_{\text{kin}} \quad (2.2.1.24)$$

From eq. (2.2.1.14) it reads for non-relativistic particles:

$$\left(\frac{dE}{dx} \right)_{\text{coll}} \propto \frac{n z^2}{v^2} \quad (2.2.1.25)$$

With $n \sim \rho$ it yields:

$$\left(\frac{dE}{dx} \right)_{\text{coll}} \propto \frac{\rho z^2 m}{E_{\text{kin}}} \quad (2.2.1.26)$$

where ρ is the density of the absorber and m the mass of the particle with charge z and kinetic energy E_{kin} . That finally leads to:

$$\bar{R} = \int_0^{E_{\text{kin}}^{(i)}} \frac{E_{\text{kin}}}{\rho z^2 m} dE_{\text{kin}} \quad (2.2.1.27)$$

and further to:

$$\bar{R} \propto \frac{(E_{\text{kin}}^{(i)})^2}{\rho z^2 m} \quad (2.2.1.28)$$

⁸ A pencil beam is a theoretical concept assuming an infinitesimal small radial extension for a particle beam.

Eq. (2.2.1.28) holds for non-relativistic particles. The last relation is known as Geiger's law of range, where the range varies as the square of the initial energy of the particle. More detailed calculations under consideration of relativistic effects show the exponent of the initial energy to be closer to 1.5.

As can be seen from eq. (2.2.1.28) the range scales inversely with the density of the absorber. This again gives rise to the introduction of a quantity $R\rho$ – the mass range.

2.2.1.2.5 Range of electrons

Due to the many deflections electrons undergo while traversing matter an average path length is defined:

$$\overline{PL} = \int_0^{E_{\text{kin}}^{(i)}} (S_{\text{tot}}(E_{\text{kin}}))^{-1} dE_{\text{kin}} \quad (2.2.1.29)$$

where $E_{\text{kin}}^{(i)}$ is the initial kinetic energy of the electron. The path length of an electron is significantly different from its (mean) range. Fig. 2.2.1.16 shows the different range definitions for charged particles. They grey area in the top indicates the path length for electrons.

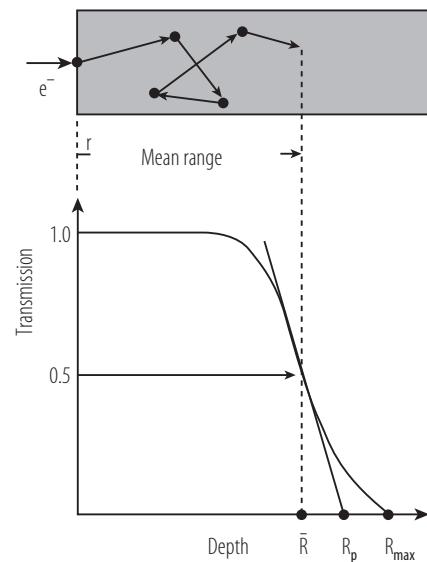


Fig. 2.2.1.16. The different range definitions for charged particles. For practical reasons a mean range \bar{R} is defined as well as a practical (extrapolated) Range R_p , both differ from each other as well as from the maximum range R_{\max} . Electrons suffer many large angle deflections resulting in a substantial difference between path length and range in the absorber as shown in the upper part of the figure (adopted from [05Kri]).

Various relations are given in the literature to estimate electron range, i.e. the maximum (mass) range of mono-energetic electrons, such as [86Tur]:

$$\begin{aligned} \rho R_{\max}(E) &= 0.412 E^{1.27 - 0.0954 \ln E}, & 0.01 \leq E \leq 2.5 \text{ MeV}, \\ \rho R_{\max}(E) &= 0.530 E - 0.106, & E \geq 2.5 \text{ MeV}, \end{aligned} \quad (2.2.1.30)$$

with E in [MeV] and ρR_{\max} in [g/cm^2]. For absorber with low atomic numbers the practical range ρR_p in [g/cm^2] for electrons beyond 1 MeV is in a very good approximation given as 50% of the numeric value of the incident electron energy in [MeV].

Radioactive isotopes emitting β -radiation exhibit continuous emission spectra of electrons or positrons where the knowledge of mono-energetic ranges is of little use. Here numeric relations are given such as:

$$D_{1/2}(\text{cm}) \approx \frac{0.108 [E_{\beta}^{\max} (\text{MeV})]^{1.14}}{\rho (\text{g}/\text{cm}^3)} \quad (2.2.1.31)$$

where E_{β}^{\max} is the maximum spectral energy and $D_{1/2}$ is the absorber depth where 50% of the electrons are stopped. Table 2.2.1.5 gives some depths $D_{1/2}$ for typical radio isotopes as used in nuclear medicine.

2.2.1.2.6 Production of charged particles

Mono-energetic electrons are almost exclusively utilized in external beam radiation therapy (EBRT) and produced by linear accelerators (LINACs, see Sect. 2.2.1.1.8). Typical electron energies are here in the range from 6 MeV to 21 MeV.

Medical applications utilizing radioactive isotopes as sources for β -radiation aim either at metabolic (functional) imaging⁹ using β^+ -emitter in Positron-Emission-Tomography (PET) or at therapeutic applications where short range β^- -radiation is advantageous¹⁰.

Table 2.2.1.5 gives a selection of the most common β -emitting radioisotopes as used in nuclear medicine.

Table 2.2.1.5. Typical β -emitting radioisotopes as used in nuclear medicine. Included are positron emitters as utilized in Positron-Emission-Tomography (PET). Note, that all isotopes given are pure β -emitters except for ^{186}Re with a γ -component from electron capture.

Isotope	$T_{1/2}$	Decay	E_{β}^{\max} [MeV]	$D_{1/2}$ [mm]
^{32}P	14.4 d	β^-	1.710	2.0
^{90}Y	64.0 h	β^-	2.27	3.6
^{169}Eb	9.5d	β^-	0.34	0.3
^{186}Re	3.7d	β^-, EC	0.98	1.2
^{11}C	20 min	β^+	0.96	1.0
^{15}o	2 min	β^+	1.72	2.0
^{18}F	110 min	β^+	0.63	0.6

Cyclotrons are used for both proton and ion acceleration. They operate at energies required to reach deep-seated tumors. To adapt the energy to the requested lesion depth, the particle usually enters an energy selection system after extraction, in the case of protons for instance simple range shifters or degraders. However, the latter introduce problems such as increased scatter and energy spread as well as significant neutron production within the degrader.

An alternative eliminating the need for energy degraders are synchrotrons. They are designed to accelerate protons and ions to the exact energy needed for therapy. In contrast to cyclotrons, however, they are pulsed machines. Maximum proton energy for therapy is ~ 250 MeV with about 10^{11} protons/spill, while maximum carbon energies range from (320 to 430) MeV/nucleon with $(0.4 \text{ to } 1.0) \cdot 10^9$ ions/spill. A spill typically lasts from 1 s to 10 s [1OPTCOG2].

2.2.1.3 Absorbed dose D and linear energy transfer (LET)

The energy loss of ionizing radiation traversing tissue is given by the stopping power, either directly for charged particles or indirectly via the secondary electrons generated in the various effects as described in Sect. 2.2.1.1. From the tissue's perspective, all or at least a fraction of this energy loss is absorbed, described as absorbed dose D :

$$D = \frac{dE}{dm}; [D] = 1 \text{ J/kg} = 1 \text{ Gray} \quad (2.2.1.32)$$

⁹ Metabolic imaging allows the detection of increased metabolism, e.g. as an indication of cancer. One way to achieve this is to tag glucose with a radioactive isotope. The glucose accumulates in the lesion (due to increased metabolism) and can be detected from outside via the emission of the attached isotope.

¹⁰ A typical example is here radio-synovectomy. The radioactive isotope is injected inside a joint to destroy the degenerated synovia. Only a few millimeters thick, the radiation burden should not extend further than absolutely necessary.

with dE as the mean energy deposition in mass element dm and Gray (Gy) as the dedicated unit. It turns out that D does not allow a complete quantification of the biological impact of ionizing radiation (see also chapter 2.3.1). The concept of *local* energy absorption leads to the Linear Energy Transfer (*LET*):

$$LET = \frac{dE}{ds}; \quad [LET] = \text{keV}/\mu\text{m} \quad (2.2.1.33)$$

Here dE is the mean energy loss of the charged particle due to (soft and hard) collisions along a path ds . In this definition *LET* is numerically equal to collision stopping power as defined in [eq. \(2.2.1.12\)](#). However, if the energy transfer in a single collision results in δ -electrons, the concept of local dose deposition is contradicted. Therefore, a limit Δ is introduced as the maximum transfer energy considered for a single collision:

$$LET_{\Delta} = \left(\frac{dE}{ds} \right)_{\Delta} \quad (2.2.1.34)$$

This restricted LET equals numerically the restricted stopping power from [Sect. 2.2.1.2.1](#).

Since LET is a criterion of ionization density, sparsely and densely ionizing radiation can be (more or less arbitrarily) separated by an LET of 3.5 keV/ μm .

As described earlier in Landolt-Börnstein Vol VIII/4 “Radiological Protection”, Chapter 2, Sect. 2.4.2.2 “Effects of dose and dose rate” [05Kau], different LETs for different radiation qualities (e.g. high energy photons or heavy charged particles) lead to a different biological impact for the *same*¹¹ physical (i.e. absorbed) dose.

2.2.1.4 Photons versus charged particles in medical applications

Radiation therapy aims at maximum tumor control at minimal side effects. Ideally, the energy released by ionizing radiation is completely deposited in the tumor with no dose applied to the surrounding tissue. However, any radiation usually needs to penetrate normal (healthy) tissue before reaching the target (i.e. the tumor or any other lesion to be treated). [Fig. 2.2.1.17](#) shows the different dose patterns with depth (depth dose curves) for different radiation qualities. These characteristics follow directly from the physics of interactions as discussed above.

The radiation dose deposited by high energy photons, after reaching a maximum, decreases exponentially with penetration depth. For a single beam the entrance dose is usually higher than the dose at the position of the target deep inside the patient's body. The higher the energy of the incident photon beam, the deeper the maximum of the depth dose distribution and the slower the decrease of dose beyond this maximum. Independent from the energy the exit dose is never zero.

The dose distribution produced by electrons at energies used in radio-oncology displays a broad maximum at a few centimeters depth. There is a distal fall-off that is moderately steep. Not shown here is the bremsstrahlung background rendering a tail beyond the finite range. The electron energies used in therapy allow treating only volumes located at relatively shallow depths.

One advantage of proton and heavy charged particle radiation compared to X-rays and electrons is their finite range and the shape of their depth dose distribution (i.e. curves). They display a very slow increase that occurs along about $\frac{3}{4}$ of the proton range. At a depth corresponding to the last quarter of the range, the dose rises considerably, reaches a maximum value and rapidly falls off to zero. This area is called Bragg peak.

For heavy charged particles, such as deuterons, helium ions, carbon ions, etc., similar depth dose distributions with a sharper Bragg peak are obtained. Here the depth dose distribution additionally displays a tail beyond the Bragg peak fall-off due to hadronic nuclear interactions, which represents about 1/10 of the peak dose value.

¹¹ It is beyond the scope of this text to discuss the problem of actually determining the absorbed dose D . As of today, practical measurements usually integrate the absorbed energy over a finite volume completely ignoring the infinitesimal definition of D . The reader is referred to Chapter 3 of the present volume.

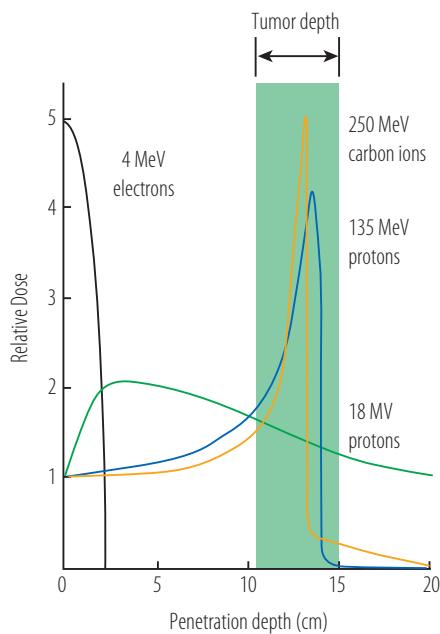


Fig. 2.2.1.17. Depth dose curves for photons, electrons, protons and heavy ions. While photons never intersect the x-axis due to their exponential attenuation (cf. Sect. 2.2.1.1.6), electrons (in a first approximation) and protons do. Heavy ions exhibit a tail due to hadronic interactions. From a clinical point of view, protons and heavy ions exhibit the way better depth dose curve with a lower entrance dose compared to the dose at tumor or lesion location. The energy is chosen such that the (Bragg) peak is positioned inside the target structure. (Drawing adopted from Siemens.com 2010).

Besides the depth dose characteristic the lateral spread (the ‘ballistics’) as another parameter for the accuracy of dose deposition shows superiority for protons and heavy charged particles. While the lateral spread is relatively large for high energy photons and electrons, protons and heavy ions penetrating tissue (equivalent material) allow a better lateral definition of dose. The higher the mass of the particle the lower the lateral spread as shown in Fig. 2.2.1.18.

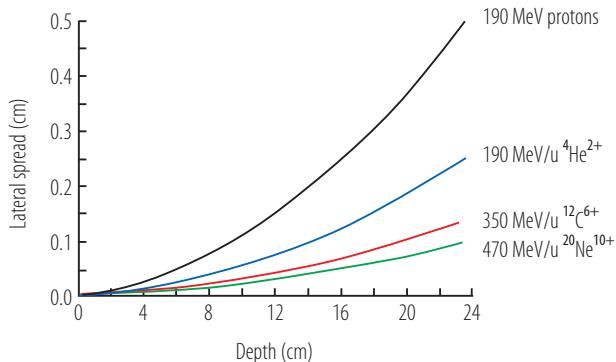


Fig. 2.2.1.18. Beam broadening due to multiple Coulomb scattering for different ion pencil beams in water. The figure shows the depth dependence of the standard deviation of the lateral spread for the different ion beams [06Schl].

2.2.1.5 Neutrons and π -mesons

Application of fast neutrons for radiotherapy of cancer has been a subject of clinical and research interest since the discovery of the neutron by Chadwick, in 1932 [94Wam]. Neutrons have been considered as advantageous over photons due to their high biological effectiveness and low oxygen dependence (see Sect. 2.3.1). However, they never convinced clinically, mainly due to the occurrence of late severe toxicities [06Jo, 10Pom]. In addition to that, to be useful at all for selected indications such as inoperable salivary gland tumors, only neutron beams produced by protons or deuterons (hitting a beryllium target) with energies greater than about 50 MeV can produce tumor control with side effects no worse than high energy photon radiation [97IAEA]. This condition results in capital costs in the order of charged particle facilities without the full-blown advantage of the latter such as superior ballistics and depth dose distribution. Most neutron therapy centers had been closed by the end of the last century. As of today,

patient numbers are rapidly decreasing in the very few fast neutron therapy centers that are still operating world-wide, almost all of them spin-offs of research institutions in contrast to the growing number of clinically dedicated charged particle centers.

Another neutron-based therapy being researched is boron neutron capture therapy (BNCT). Here, a neutron capture agent with very high thermal-neutron capture cross section (^{10}B) is selectively taken into the malignant tissue following the administration of a suitable boron delivery agent. The treatment volume is then exposed to thermal neutrons interacting with the ^{10}B , which is supposed to be present in the malignant cells only. The subsequent boron-neutron interactions produce alpha particles and lithium ions. They deposit their energy locally, i.e. within the malignant cell, ideally leading to a high overall tumor control. Applied mostly to incurable brain tumors, the results are controversially discussed throughout literature [08Li].

Negative π -mesons have also been used for therapy. In addition to the increased ionization density in the Bragg peak region, the pion is absorbed by the nucleus of the atom where it stops causing it to disintegrate with the release of a substantial amount of additional energy. This extra "star dose" adds substantially to the biological damage. Pions are harder to use, however, because of the difficulties of producing high-enough dose rates and for ballistic reasons (mass of the pion is only 7% that of a proton). The lower mass causes increased multiple scattering and range straggling, so the precision of placement of a pion beam is quite worse than that for a proton. The fact that pions are secondary particles makes them very expensive to produce. They require very high fluxes of protons at energies above 600 MeV to strike a production target, and a highly sophisticated transport channel to separate out the pions of the desired energy from all the contaminants produced in the target [93Alo]. Nevertheless, three centers around the world have treated more than 1000 patients with pions. Pion treatments started at Los Alamos (USA) in 1974, followed by TRIUMF (Vancouver, B.C. Canada) and PSI (Villigen, Switzerland)). Los Alamos stopped in 1982, TRIUMF and PSI gave up pions around 1990. After then both facilities focused on proton therapy.

2.2.1.6 Interaction modeling in radiation therapy treatment planning

As stated throughout the text above, both indirectly and directly ionizing radiation is used in current radiation oncology treatment regimen. There is a wide variety of different modalities to deliver radiation such as Particle Beams or External Beam Therapy (EBRT), both with their modern applications such as the different techniques of Intensity Modulated Radiation Therapy (IMRT). Others, such as Brachytherapy (see chapter 2) or Intra-Operative Radiation Therapy (IORT) are amending this pool of options. Different modalities can be combined for a particular indication, e.g. prostate cancer is often treated employing EBRT and Brachy-therapy. At the same time, modalities often compete for the same indication as, again, in case of prostate carcinoma treated either with IMRT or permanent seed implantation, a sub-specialty of Brachy-therapy. To further elaborate on these techniques and their indications in radio-oncology is beyond the scope of this text since, as of today, there is no strict link between indication (group) and treatment modality.

Prior to any application of ionizing radiation within an oncology treatment regimen a computer based simulation of the intended treatment is performed, usually referred to as Treatment Planning. Here the dose distribution eventually achieved in the patient is calculated as a function of irradiation parameters¹². A 3-D set of patient data acquired by different imaging methods such as Computed Tomography (CT), Magnetic-Resonance-Imaging (MRI), Positron-Emission-Tomography (PET) and/or a combination of the latter¹³ serves as the basis for the simulation. While one area of intense research is the automatic optimization of irradiation parameters to achieve the desired dose distribution (cf. [10Bor, 09Bor, 07Ulrl]), the calculation of the dose distribution itself due to pre-defined irradiation parameters also constitutes a

¹² Irradiation parameters depend on the modality used and vary widely. Simple examples are beam directions in EBRT or seed distributions in the permanent seed implantation (PSI) for prostate cancer.

¹³ It should be noted, that a set of CT-data is always necessary since they contain the electron densities necessary for dose calculations. Other image modalities are usually added for improved delineation of the dose target region and the areas to be saved from radiation burden.

crucial link in the chain of radiation therapy planning. As of today, models used here are still a trade-off between calculation speed and accuracy.

Simple dose calculation models are based on a large pool of pre-measured data (so-called look-up tables) specific for the radiation quality used. The result for a concrete set of irradiation parameters is then the superposition of interpolations from different look-up tables. All commercial Treatment Planning Systems (TPS) offer this option. However, the considerable speed of this method is paid by limited accuracy and the restriction to only simple configurations of irradiation parameters.

Therefore, most TPS today work with Pencil Beam methods ([10Hus](#)) or Collapsed Cone approaches ([07Has](#)). Here the dose is calculated by superposition of so-called dose kernels. The latter are 3-D data sets derived from first radiation interaction principles as described in the sections above. A Monte-Carlo (MC) simulation is used to build the kernels specifically for the radiation (quality and spectrum) used. The accuracy of these methods depends, besides the kernels, obviously on the modelling of the radiation fluence (outside and inside the patient) since it is the fluence that ‘scales’ the impact of a kernel at a specific location. The superposition of the kernels and the problem of fluence modelling slow down calculation speed depending on the accuracy desired.

The ultimate dose calculation is based on a MC-simulation of the complete radiation transport and dose deposition process. There are algorithms for photons, electrons and protons^{[14](#)} (cf. [[10Fra](#), [07Sec](#)]) that are adapted to clinical use. However, the calculation time is still such that MC-simulation based treatment planning is a rare exception in current clinical routine. Nevertheless, with the increase of calculation speed due to improved computer hardware MC-based dose calculation will soon become the (gold) standard in radiation treatment planning.

While for high energy photons energy dose distributions obtained/predicted by methods as described above are taken as equal to the biological impact, heavy charged particles require the additional consideration of their increased biological effectiveness. A simple approach here is the notation of dose in cobalt-equivalent Gray [[10Mac](#)]. More subtle approaches of biological treatment planning are currently being introduced into clinical practice (cf. [[09Kro](#),[10Sch](#)]).

2.2.1.7 Summary

Ionizing radiation has always been part of human environment. Life on earth is adapted to the naturally occurring exposure levels. As of today, the biological impact mechanisms of ionizing radiation are fairly understood. In conjunction with epidemiological and systematically derived data there is evidence of exposure thresholds beyond which radiation exposure becomes detrimental to health and/or where energy transport is sufficient to use it therapeutically.

Ionizing radiation today is being used extensively in medical applications where in conjunction with a reliable radiation protection, the benefit far outweighs possible health detriments.

An inevitable prerequisite to biological evaluations is a thorough understanding of the ionizing radiation physics. Different interaction mechanisms in matter give rise to the clear distinction between indirectly and directly ionizing radiation. The vast majority of natural and man-made sources are emitting photons, either solely or partly in a multi-step process. Today, photons as indirectly ionizing radiation are the best understood radiation quality. [Sect. 2.2.1.1](#) summarized their physics and production under the perspective of medical applications as well as radiation protection.

Charged particles as directly ionizing radiation, particularly heavy charge particles (protons and beyond) are of increasing importance in radiation oncology. Here the interaction with matter results in an obviously more suitable energy dose distribution than (even high energy) photons do (cf. [Sect. 2.2.1.4](#)).

However, charged particles still have to prove the translation of physical superiority over photons into better clinical results. From a radiation protection point of view, charged particles at energies as encountered in most applications are easily shielded due to their limited range. Physics and production of charged particles is summarized in [Sect. 2.2.1.2](#).

¹⁴ For heavy ions, MC-based calculations exist. However, heavy ions need to be treated differently in the whole context since their significantly increased biological effectiveness requires a biological treatment planning with physics only a necessary but not sufficient step in this process.

Early particle therapy also included π -mesons and neutrons. Both did not live up to the clinical expectations and are found only occasionally in the clinic today or, in the case of π -mesons, have vanished completely. Sect. 2.2.1.5 gives a brief overview of the pertinent history.

This Sect. 2.2.1 focused on the impact of the absorbing matter onto the radiation impinging the absorber, i.e. on the radiation transport as a function of the matter interacting with it. Henceforth the perspective switches. The following Sect. 2.3.1 will focus on the biological impact of ionizing radiation, i.e. on the biological processes triggered by the impinging radiation.

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2.2.2 Non-Ionizing Radiations

Radiation and Biological Effects Kinds of Radiation

J.H. BERNHARDT

2.2.2.1 Introduction

While ionizing electromagnetic radiation clearly refers to all radiation having individual photon energies sufficient to ionize atoms, an absolute statement regarding non-ionizing radiation (NIR) is difficult. Normally, lower-energy photons (i.e., ≤ 12 eV) have insufficient energy to release a bound electron from an atom. However, in solid-state matter, electrons can be released by much lower photon energies. Furthermore, power laser beams can be focused to produce plasma-ionized matter. In general, for purposes of health protection, electromagnetic NIR can be subdivided into a number of wavelengths (λ) or frequency (f) ranges (see [Table 2.2.2.1](#)):

- ultraviolet radiation, $100 \text{ nm} \leq \lambda \leq 400 \text{ nm}$ (optical radiation);
- visible radiation, $400 \text{ nm} \leq \lambda \leq 760 \text{ nm}$ (optical radiation);
- infrared radiation, $760 \text{ nm} \leq \lambda \leq 1 \text{ mm}$ (optical radiation);
- laser radiation at different wavelengths of the optical radiation;
- radiofrequency radiation including microwaves, $100 \text{ kHz} \leq f \leq 300 \text{ GHz}$ corresponding to $3000 \text{ m} \geq \lambda \geq 1 \text{ mm}$,
- extremely low frequency (ELF) fields ($f \leq 300 \text{ Hz}$), in practice mainly power frequencies of 50-60 Hz.

From a pragmatic point of view, magnetostatic and electrostatic fields are also dealt within the framework of NIR. The field of NIR comprises also pressure waves such as ultrasound and infrasound including airborne ultrasound and infrasound. That is on either side of the audible frequency range (20 Hz – 20 kHz).

Such radiation can be emitted continuously or intermittently and the modulations that affect the frequency, amplitude, or pulse can modify some of the health aspects. Highly directional beam sources such as lasers may present special problems. A fundamental distinction between electromagnetic fields and ultrasound is the necessity for acoustic vibrations to have a physical medium through which to propagate. They cannot be transmitted through a vacuum, as can the electromagnetic radiations and fields.

The electromagnetic environment consists of natural fields and man-made electromagnetic fields that are produced either intentionally or as by-products of the use of electric devices.

The natural electromagnetic environment originates from terrestrial and extraterrestrial sources such as electrical discharges in the earth's atmosphere and radiation from sun and space. This natural background is orders of magnitude below local field levels produced by man-made electromagnetic sources considered here. The everyday use of devices and systems emitting electromagnetic fields is continuously increasing. Sources generating high levels of electromagnetic fields are typically found in medicine and at certain workplaces. Medical devices used for magnetic resonance imaging, diathermy, hyperthermia, various kinds of radiofrequency radiation (RF) ablation. Surgery and diagnoses may cause high levels of electromagnetic fields at the patient's position or locally inside his body. In addition, some of these medical applications may cause high fields at certain workplaces.

For broadcasting high RF power is generally required to maximise the area of coverage. Close to the antennas electric field strengths can reach several hundred volts per meter. Even higher values can be found close to occupational sources used for processing of various materials by heating and sometimes by formation of plasma discharge in the material. In many such applications RF-safety problems arise because RF-power is high and it may be difficult to enclose the field-generating electrodes and processing space inside a good electro-magnetic shield. Sources used by the general public e.g. for

wireless communication, data transmission or food processing generates comparably much lower fields at the position of the user. But this may also depend on the behaviour of the user especially concerning the distance of the source.

Mobile telephony networks cause on average low levels of electromagnetic fields in areas accessible for the general public. Cell phones, however, might cause high peak levels of exposure during use.

Electronic article surveillance (EAS) systems and radio frequency identification devices (RFID) operate at many different frequencies within the RF band. Inside some EAS gates electromagnetic fields could get close to the limit values. In general these systems cause only low fields in the environment.

Radars produce high power main beams only a few degrees wide and usually not accessible during operation. In addition radar antennas typically rotate and signals are pulsed, leading to a reduction in average exposure.

Electromagnetic radiation with wavelengths in the range between 100 nm and 1 mm is widely termed optical radiation. Ultraviolet radiation (UV) is in the wavelength range between 100 nm and 400 nm, light (or visible radiation) from 400 to 760 nm and infrared (IR) radiation from 760 nm to 1 mm. While these spectral bands are useful "short-hand notations" for discussing the photobiological effects of optical radiation, the predominant biological effects have less sharply defined spectral limits. Because there is little energy emitted by conventional light sources in the far-infrared (IR-C range), only lasers in this spectral region pose potential hazards.

Table 2.2.2.1. Ranges of frequency and wavelength for some types of electromagnetic radiation

Type of radiation	Frequency range	Wavelength range
Ionizing > 3000 THz	<100 nm	
Ultraviolet (UV) (non-ionizing part)	3000–750 THz	100–400 nm
– UV-C	3000–1070	100–280
– UV-B	1070–952	280–315
– UV-A	952–750	315–400
Visible radiation (light)	750–385 THz	400–780 nm
Infrared (IR)	385–0.3 THz	0.78–1000 μm
– IR-A	385–214	0.78–1.4
– IR-B	214–100	1.4–3
– IR-C	100–0.3	3–1000
Lasers	1500–15	0.2–20
– Class 1	non-risk laser devices	
– Class 2	low-risk, low-power laser devices	
– Class 3a	low-risk, medium-power laser devices	
– Class 3b	moderate-risk, medium-power laser devices	
– Class 4	high-risk, high-power laser devices	
Radiofrequency (RF)	300 GHz–0.1 MHz	1 mm–3000 m
– Extremely high frequency (EHF)	300–30 GHz	1 – 10 mm
– Super-high frequency (SHF)	0–3	10–100
– Ultra-high frequency (UHF)	3–0.3	100–1000
– Very high frequency (VHF)	300–30 MHz	1–10 m
– High frequency (HF)	30–3	10–100

Type of radiation	Frequency range	Wavelength range
- Medium frequency (MF)	3–0.3	100–1000
Low frequency (LF)	300–30 kHz	1–10 km
Very low frequency (VLF)	30–3	10–100
-	3–0.3	100–1000
Extremely low frequency (ELF)	<0.3 kHz	>1000 km

The special optical properties of lasers vary significantly from those of conventional broad-band optical sources, and so they are treated separately. Unlike lasers, which generally emit at one wavelength, spectroradiometric data, measurements and geometrical factors must be considered when evaluating potential hazards of conventional light sources.

Optical radiations including lasers are widely used for technical and medical applications. Examples are illumination, photography, chemical processes, printing, arc welding, arc cutting, sterilization and laser material processing in industrial processes, medical applications include diagnostic purposes in dentistry, UV phototherapy (PUVA), photo chemotherapy, photodynamic therapy and IR hyperthermia. For medical applications lasers are used in general surgery, in ophthalmology, gastroenterology, dermatology and urology.

In the next sections the different types of non-ionizing radiations will be treated as follows, whereby the order of the electromagnetic field ranges is within the sequence of increasing frequency:

- 2.2.2.2 Ultrasound,
- 2.2.2.3 Static and slowly varying electric and magnetic fields,
- 2.2.2.4 Time varying electric and magnetic field of frequencies less than 100 kHz,
- 2.2.2.5 Electromagnetic fields of frequencies above 100 kHz,
- 2.2.2.6 Optical radiation and lasers.

The sections will be treated – as far as possible - within a homogeneous structure:

- Physical characteristics,
- Sources,
- Technical and medical applications,
- Interaction mechanisms,
- Summary

2.2.2.2 Ultrasound

2.2.2.2.1 Introduction

Ultrasound is a form of mechanical energy that has found increasingly widespread application over the past 50 years. Many of its uses entail exposure of human beings, either incidentally or, as in the case of medical applications of ultrasound, as an essential part of the procedure. The fact of such exposures inevitably raises the question of the possible existence of any corresponding hazard to the individual exposed.

The physical nature of ultrasound, the physical laws that determine its behaviour, and the principal methods that are available for measuring the characteristics of ultrasound fields are first described, after which the main practical applications of ultrasound are considered from the point of view of their potential for exposure of humans to ultrasound fields. With describing the biophysical mechanisms, the evidence that exists for the induction of biological change is considered. For further reading see the reviews, for example [83NCR, 93Zis, 07Bar]. The Sections below are mainly based on these reviews.

2.2.2.2 Physical characteristics

The term ultrasound refers to the set of mechanical vibration phenomena that occur in the frequency range above the upper frequency limit for human hearing, which may be taken as lying at about 16 kHz. The upper frequency limit is set by practical considerations in the generation of mechanical vibration and, with present technology, lies at about 100 GHz. The practical upper limit for concern as to human exposure is about 20 MHz.

In contrast to electromagnetic radiation, ultrasound is fundamentally related to the existence of a material medium and cannot be sustained *in vacuo*. The physical nature of the vibration phenomena that may occur is strongly dependent on the phase of the medium concerned - solid, liquid or gas - and will also be greatly influenced by vibration frequency over the very wide spectrum referred to above (16 kHz-100 GHz).

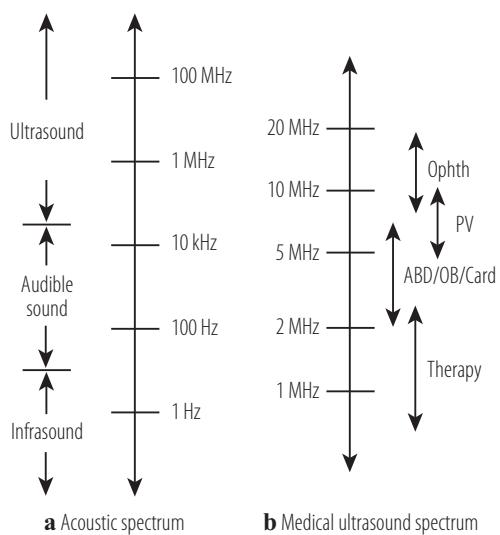


Fig. 2.2.2.1. Acoustic (a) and medical ultrasound spectrum (b).

The generation of ultrasound vibrations in a medium entails the existence of fluctuations in the values, at different points in the medium, of certain of its physical parameters. In particular, in considering any very small particle/volume of the medium (small in relation to the wavelength) it will generally be the case that fluctuations occur in the displacement s (a vector quantity) of the particle/volume from its resting position. Similar fluctuations will occur in the instantaneous values of the particle velocity v , the particle acceleration a , and the instantaneous pressure p . In the particular case that the fluctuation is sinusoidal in form, the field parameters s , v , a , p will exhibit maxima whose magnitudes S , V , A , P are referred to as displacement, velocity, acceleration and pressure "amplitudes". For certain simplified situations, such as that of a linear plane progressive wave disturbance, these four parameters are simply related and a measurement of the spatial distribution of any one of them may serve to characterize completely, or to map, the sound field.

Sound speed

An ultrasound disturbance that is induced in any medium will travel away from its source with a certain speed that is characteristic of the medium. It is this speed c which, taken together with the frequency f of the vibration induced, determines the wavelength $\lambda (= c/f)$ of the waves that are propagated. For example, the propagation speed of ultrasound in most human soft tissues is approximately 1500 m s^{-1} , so that an induced frequency of 1 MHz leads to a wavelength of 1.5 mm. It follows from the fundamental physical principles of diffraction that ultrasound diagnostic procedures carried out, for example, at frequencies around 3 MHz have the potential for providing anatomical detail of the order of 0.5 mm. Values of the sound speed for some other media are given in [Table 2.2.2.1](#). It may be noted that in unbounded solids, ultrasound may be propagated in two different modes: as compressional (longitudinal) waves, as in the

case of liquids and gases, or in the form of shear (transverse) waves. The propagation speed for the two modes is generally different.

Transmission through interfaces

A second property of a medium, which is of fundamental importance in determining the extent to which ultrasound energy is transmitted through an interface separating two continuous isotropic media, is its characteristic acoustic impedance Z , which is defined as the product ρc , where ρ is the density of the medium and c is the speed of sound in this medium. The quantitative role of characteristic acoustic impedance can be seen from the expressions (for normal incidence of an ultrasound wave on a plane interface) for the fractions P_t/P_i and P_r/P_i respectively, of the incident sound pressure amplitude P_i transmitted through P_t , and reflected back from the interface:

$$P_t/P_i = 2Z_2/(Z_1 + Z_2) \quad (2.2.2.1)$$

$$P_r/P_i = (Z_2 - Z_1)/(Z_2 + Z_1) \quad (2.2.2.2)$$

where P_i is the acoustic pressure amplitude in the wave incident on the interface, P_r is that in the reflected wave, P_t is that in the transmitted wave, and Z_1 and Z_2 are the characteristic acoustic impedances of the media on either side of the interface.

Representative values of the characteristic acoustic impedance are listed in [Table 2.2.2.2](#). From inspection of these values and the above relationships it will be seen, for example, that an interface between air and human tissue permits transmission of only some 3% of the incident pressure (or 0.1 % of incident intensity, which varies as the square of pressure), the remainder being, reflected. This is an important fact because of its bearing on the possibilities for exposure of humans by airborne ultrasound or, more generally, by sources shielded by an air gap. Strong, reflections also occur at bone/soft tissue interfaces, and this constitutes another of the important limitations on the accessibility of human anatomical regions to ultrasound investigation and exposure.

Table 2.2.2.2. Ultrasound properties of various media

Medium	Sound speed c (compression wave) [m s ⁻¹]	Characteristic acoustic impedance z [kg s ⁻¹ m ⁻²]	Attenuation coefficient α at 1 MHz [neper cm ⁻¹]	Frequency dependence of attenuation coefficient, n
Air	330	4.50×10^2	15	2
Water	1500	1.50×10^6	0.02	2
Liver	1540	1.54×10^6	8	1.2
Fat	1440	1.40×10^6	5-25	1.4
Blood	1520	1.61×10^6	2.3	1.3
Skeletal muscle	1520	1.70×10^6	10-35	1.1
Skull bone	3360	6.99×10^6	115	1.7

Attenuation

A third property of a medium that needs to be considered is its ability to attenuate a propagating wave. The pressure p_x in a plane progressive sound wave of initial pressure amplitude P_0 , after travelling a distance x in any attenuating medium, is described by the relationship:

$$P_x = P_0 e^{-\alpha x} \quad (2.2.2.3)$$

where α is termed the attenuation coefficient of the medium (in neper per unit propagation distance) and e is the base of natural logarithms. If the decibel notation is used, the corresponding attenuation coefficient in dB per m is equal to 8.68α ([Table 2.2.2.2](#)).

The attenuation phenomenon is important in the present context from two points of view. In the first place, at least part of the attenuation will in general be due to a process of absorption, in which

propagating energy is transferred to the medium in the form of heat, with the possibility of causing permanent modification as a result of the corresponding rise in temperature. Second, the attenuating property of one region of a structure will evidently affect the extent to which propagating energy penetrates to deeper regions. An important point to note is that ultrasound attenuation by a medium generally increases with increasing frequency.

It should be noted that attenuation in human soft tissues is fairly high. This corresponds, on the one hand, to the effective interactions that are exploited in ultrasound therapy (referred to below) and, on the other, to the existence of a practical upper limit of about 20 MHz.

As the ultrasonic pressure wave propagates through a medium, the elements are subjected to a positive and a negative pressure. During the positive phase of the wave the elements are compressed, and during the negative phase they are pulled apart. This changes the density of the medium so that it is a little denser during the positive phase and less dense during the negative phase of the pressure wave. The velocity of propagation of the ultrasonic wave changes with the density of the medium. Thus, the positive component of the wave propagates slightly faster than the negative components. At sufficiently high pressure, this effect can lead to a gradual distortion of the wave whereby the positive components of the wave bunch up, leading to the development of a steep positive wave-front, referred to as a shock wave [94Nau]. The negative components also suffer some distortion and the negative portion of the wave is compressed and stretched out by a small amount.

Absorption and heating

In most materials, including human tissues, a substantial part of the process of attenuation is that corresponding to the direct conversion of acoustic energy to heat, and generally referred to as absorption. Quantitatively it can be written that:

$$\alpha_f = \alpha_{af} + \alpha_{sf} \quad (2.2.2.4)$$

where α_{af} and α_{sf} are the absorption and scattering coefficients of the medium respectively, at frequency f , and where the process of scattering is one in which incident energy is deflected out of the volume of interest without conversion to heat.

The corresponding generation of heat within a small volume (the absorbed power) is given, in a plane travelling wave by:

$$dh/dt = 2I \alpha_{af} [W m^{-3}] \quad (2.2.2.5)$$

where I is the ultrasound intensity incident on the volume. Thus, in the absence of any process of removal of heat, the resulting rate of rise of temperature in the volume will be:

$$dT/dt = 2I \alpha_{af} C_v^{-1} [K s^{-1}] \quad (2.2.2.6)$$

where C_v is the heat capacity of the medium per unit volume: approximately $4.2 \times 10^6 J m^{-3} K^{-1}$ for human soft tissues.

Using the data in [Table 2.2.2.2](#) and making the conservative assumption that most attenuation is due to absorption, it follows that the maximum initial rates of temperature rise in tissue consequent on exposure to intensity I [$W m^{-2}$] at frequency f [MHz] will be of the order of $3.8 \times 10^{-6} I f^{1.2} K s^{-1}$ for soft tissues, and approximately $55 \times 10^{-6} I f^{1.7} K s^{-1}$ for bone (assuming heat capacity similar to that for soft tissue).

In practice, there will almost always be substantial heat removal from the exposed volume, which will set a limit to the maximum temperature.

2.2.2.2.3 Pulsed mode of operation

The majority of ultrasonic diagnostic equipment functions in a pulsed mode [93Kre]. The principles will be treated within the corresponding sections of Landoldt Börnstein Volume VIII/7B.

Pulse duration is generally given in units of μs . The time interval between pulses is known as the pulse repetition period and, as it lasts longer, is generally given in units of ms. The pulse repetition

frequency and the duty ratio are two other commonly quoted parameters. The former is the inverse of the pulse repetition period and the latter is the ratio of the pulse repetition period over the pulse duration.

The peak positive value of the pressure in the wave is known as the peak compressional (p^+) pressure and the peak negative pressure as the peak rarefactional pressure (p^-). In an undistorted wave these two values are equal. In a fully developed shock wave, p^+ can be up to three times greater than p^- .

Two important physical properties to describe ultrasound exposure are power and intensity. Power is defined as the rate at which energy is transformed from one form into another, such as from mechanical into thermal form: Power = energy / time.

Power is specified in watts, whereas energy is specified in joules. It is an important exposure parameter in that it is a major determinant of the ability of the ultrasonic beam to heat tissue.

Intensity is the spatial concentration of power: Intensity = power / area. Intensity is specified in W m^{-2} or some derivative thereof, such as mW cm^{-2} . Intensity is proportional to the square of the pressure in the ultrasonic wave, and can therefore be calculated if the pressure is known.

Intensity is a parameter that is space-and time-dependent. The following illustrate three space-dependent definitions of intensity. Close to the transducer, before the ultrasonic beam is able to form its shape, the beamwidth is equal to the physical dimensions of the transducer. The average intensity obtained by dividing the power by the surface area of the transducer accurately reflects the uniform intensity distribution across the beam at the surface of the transducer.

As one moves away from the transducer, the ultrasonic beam forms into the shape determined by the frequency and the physical dimensions of the transducer. The distribution of the energy across the beam becomes non-uniform and there is concentration of energy on the axis. The average intensity obtained by dividing the power by the beamwidth grossly underestimates the axial intensity, typically by a factor of 3-5. As a bioeffect is most likely to occur where there is maximum intensity, the axial rather than the average intensity is the parameter of greater interest at distances away from the transducer.

The peak spatial intensity is the maximum intensity in the beam-pattern. In water it corresponds to the axial intensity at the focal distance. In tissue, because of attenuation, it may occur at a distance closer to the transducer.

The time-dependence of intensity is best illustrated by considering the ultrasonic pressure wave. Intensity is proportional to the square of the pressure; thus, the intensity in the wave has only positive values which range from zero to some peak temporal value. Indicated in Fig. 2.2.2 are some of the temporal definitions for intensity. The spatial peak temporal peak intensity (I_{SPTP}) is the peak value of the intensity in the waveform. The maximum intensity (I_m) is the intensity content in the largest single half-cycle in the waveform. The spatial peak pulse average intensity (I_{SPPA}) is the intensity content in the whole pulse, and the spatial peak temporal average intensity (I_{SPTA}) is the spatial peak pulse average intensity I_{SPPA} divided by the duty ratio. These terms are described in more detail in other publications on exposimetry [93Zis].

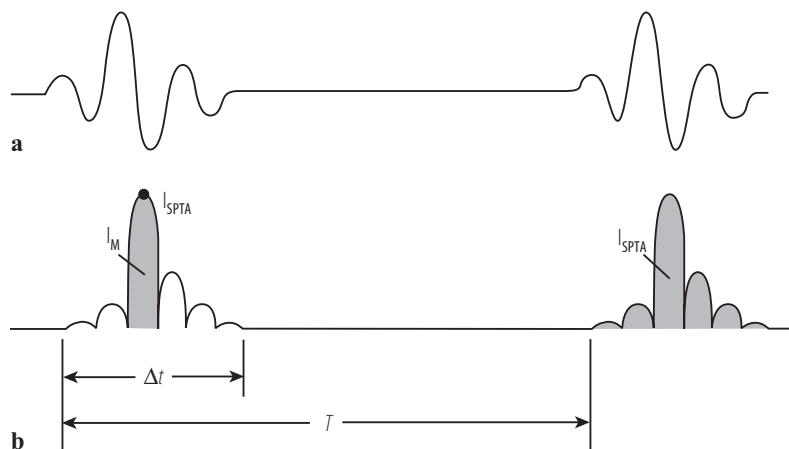


Fig. 2.2.2. (a) Ultrasonic pressure pulse and (b) intensity waveform. I_{SPTA} , spatial peak temporal average intensity; I_{SPPA} , spatial peak pulse average intensity; I_m , maximum intensity. Duty ratio = $T/\Delta t$; $I_{\text{SPTA}} = I_{\text{SPPA}} \Delta t/T$. From [98Bar].

2.2.2.2.4 Medical applications

The medical applications will be treated in detail within the corresponding sections of Landoldt-Börnstein Volume VIII/7B. The following text gives some general information.

There is a large area in a variety of medical specialties including cardiology, vascular disease, ophthalmology, internal medicine gynaecology and obstetrics. The predominant medical diagnostic application of ultrasound is in the frequency range 2-10 MHz, with occasional extension to 20 MHz for some eye examinations. Ultrasound irradiation falls into two categories – pulsed fields and continuous wave – the latter being used for Doppler frequency-shift identification of moving structures and for blood flow velocity measurement. Almost all accessible anatomical sites are liable to examination; those for which the technique has so far proved to be of most clinical value (e.g., the abdomen, pregnant uterus, heart and major vessels) are the most commonly irradiated. Accessibility is impaired to some extent by bone and particularly by air spaces; thus the inflated lung, and, in some patients, gassy bowels constitute impenetrable barriers.

The capability of high ultrasound intensities both to heat and to destroy living tissue allows its use for *physiotherapy*. There is little doubt that ultrasound used in physiotherapy constitutes the most widespread source of ultrasound irradiation of humans at power levels sufficient to cause biologically significant rises in tissue temperature. This is an old established technique in which ultrasound beams in the approximate frequency range 0.75-3.0 MHz are directed into a patient with objectives such as stimulation of capillary circulation, relief of pain and acceleration of tissue regeneration [92Sta]. Some therapeutic effects of ultrasound may well be due to biophysical mechanisms other than those related to temperature rise, for example the observed acceleration of healing in chronic varicose ulcers. A number of other effects, including cavitation, may occur simultaneously with local tissue heating. Many of the treatment methods currently in use involve such low spatial average temporal average intensities that they are unlikely to be accompanied by temperature rises above that occurring, normally as a result of diurnal variations. The exposures are limited under international standards [96IEC].

Prolonged exposure to temperatures in the range 42-45°C is lethal to mammalian cells, and this is used in the cancer therapy known as *hyperthermia*. In this technique, the tumour is heated to these levels, while an attempt is made to leave the surrounding normal tissue undamaged. Local heating can be achieved with ultrasound, although microwave and radio-frequency methods are also used. Plane transducers are used for the heating of superficial volumes, whereas focused transducers or multi-transducer arrays are used for deeper seated tumours. It has been shown that some human tumours respond to treatment by heat alone, but most benefit is obtained when heat is combined with either radiotherapy or chemotherapy.

A considerable variety of ultrasound techniques is used in connection with *surgical procedures*. These fall into two groups: (a) those in which ultrasound vibration is used to increase or modify the effectiveness of a surgical instrument; and (b) those in which an ultrasound beam is generated and it used to effect some surgical action. Few of these techniques in either group seem yet to have achieved more than limited, experimental use; the one with the most widespread application is probably the use of miniature beams in the treatment of Ménière's disease of the inner ear. Focused ultrasound surgery has been used with some success to treat glaucoma [85Col]. Devices using 40 kHz ultrasound are now employed frequently in procedures for cataract removal. A small tip, rather like that of a hypodermic needle, is introduced into the anterior chamber of the eye, placed in contact with the lens, and then set into longitudinal oscillation. This causes emulsification of the cataract body which, especially for mature cataracts, facilitates its removal by aspiration [73Kel].

Tools driven by ultrasound have been used for many years for the disintegration of renal calculi [78Fah]. An important application is the fragmentation of kidney stones using shock waves (extracorporeal shock wave lithotripsy). In this procedure, patients are carefully positioned in a large water bath such that the kidney stone lies at one focus of an ellipsoidal mirror. A spark discharge is generated at the other focus. The amplitude of the shock wave is at its maximum at the stone, and causes its disintegration. Pressure amplitudes of 14 MPa have been reported at the focus [86Sau, 89Col]. Little is known about the effect of the transit of such shock waves through soft tissues.

Applications in dentistry are widely used for removing calculus and other deposits from teeth. Suitably shaped tips are mounted on the end of a vibrating rod in the ultrasound dental handpiece. The

vibrating tip is slid over the tooth surface, applying a shearing force to the calculus layer being removed [68Ewe]. Dental handpieces operate at frequencies of about 18-40 kHz, with a vibration amplitude of about 0.01-0.025 mm. Water is sprayed against the vibrating tip for cooling purposes and the cavitation effects in the liquid being sprayed reinforce the mechanical effect of the vibrations in removing deposits. Estimated total power output levels of 10 W have been reported.

2.2.2.2.5 Acoustic parameters used to describe ultrasound exposure

Introduction

In the past many acoustic parameters have been used to describe acoustic exposure. Initially, owing to limited availability of measuring equipment, only the power output or the average intensity was used to describe the exposure. An example of this approach was the early American Institute of Ultrasound in Medicine (AIUM) recommendation for an upper limit of 100 mW cm^{-2} for the average intensity.

The development of miniature hydrophones gave investigators the ability to measure the ultrasonic pressure wave quantitatively and calculate the intensity in the various parts of the wave. This capability allowed, for example, the Food and Drug Administration regulatory authority in the USA to specify permissible levels of acoustic output for cardiac, vascular, ophthalmic and fetal applications expressed in terms of three parameters for intensity, namely I_{SPTA} , I_{SPPA} and I_m [93FDA].

With time, it was recognized that these three intensities did not allow adequate comparison of the ability of different ultrasonic beams to produce a potential bioeffect. The emphasis has since shifted to the use of parameters that more closely permit such comparison. This has resulted, for example, in the introduction by the AIUM of on-screen real-time display of non-acoustic indices such as the thermal and the mechanical index to describe the output being generated by the equipment.

Currently the acoustic output of diagnostic equipment is specified in terms of the maximum value of four acoustic parameters of the ultrasonic field. The first two parameters, the acoustic power output and the spatial peak temporal average intensity I_{SPTA} , relate to the ability of the ultrasonic beam to heat tissue and so potentially cause a thermally induced bioeffect. The third and fourth parameters, the spatial peak pulse average intensity I_{SPPA} and the peak negative pressure p^- describe the intensity content and peak pressure value of the transmitted pulse waveform and relate to the potential ability of the pulse to cause a cavitationally induced bioeffect.

It is recognized that the first two parameters do not permit calculation of the actual temperature rise in tissue. For this it is also necessary to know the value of other acoustic field parameters such as beamwidth and frequency, and of acoustic and physical properties of tissues such as attenuation and perfusion. Similarly, the last two parameters are insufficient to predict possible occurrence of cavitation in tissue. Indeed, the theory for cavitation in tissue has not as yet been formalized, and there are many uncertainties in our understanding of the requirements necessary for the development of cavitation in biological liquids and in soft tissue. Nevertheless, it is generally agreed that these four parameters allow a qualitative assessment and comparison of output of different equipment.

Acoustic power output

When an ultrasonic beam, propagating in water, impinges on an absorbing target such as a piece of butyl rubber, the target experiences a radiation pressure force that is proportional to the power contained in the beam. The radiation pressure force is relatively small; a beam with a power content of 1 W exerts on the target a weight of 0.067 g. The power output of modern equipment ranges from 1 to 500 mW, and is capable of generating a radiation pressure weight ranging from 0.067 to 0.0335 g. Although small, this radiation pressure weight can be measured by attaching a rubber target to the measurement arm of a sensitive analytical balance and noting the increase in weight of the target when the transducer is energized.

In M-mode and pulsed Doppler, the ultrasound beam is stationary and the size of the absorbing target needs to be just a little larger than the beamwidth to measure the energy content in the beam. In B-mode and colour Doppler, the ultrasound beam is scanned to form the image. In this application the size of the target needs to be large, to ensure that all of the beams forming the image are intercepted and contribute to the power output measurement.

Spatial peak temporal average intensity

I_{SPTA} is measured by a calibrated hydrophone. These hydrophones are very small (typically 0.5 mm in diameter) transducers made from a piezoelectric plastic film.

The intensity waveform is obtained by squaring this waveform. The resultant waveform is shown in Fig. 2.2.2.2. The spatial peak pulse average intensity is obtained by integrating this waveform, i.e. calculating the area under the waveform. Fig. 2.2.2.2 shows that the ultrasonic pulse has certain duration t and that the pulses are repeated every T seconds. The spatial peak temporal average intensity is obtained by multiplying the spatial peak pulse average intensity by the pulse duration t and dividing it by the pulse repetition time T . Most well equipped laboratories have a small computer attached to the hydrophone to allow these calculations to be performed automatically.

Spatial peak pulse average intensity

I_{SPPA} illustrated in Fig. 2.2.2.2 describes the intensity contained in each pulse. This parameter is required for the derivation of the spatial peak temporal intensity. As the likelihood of the onset of cavitation is related to the spatial and temporal concentration in intensity, it is also used on its own merit as a descriptor for the potential of the ultrasound exposure to cause a cavitationally induced bioeffect.

There are many gaps in our knowledge of the requirements necessary for development of cavitation in biological media. As such knowledge is derived, it could well be that this parameter will be superseded or be complemented by other parameters which would be better reflections of the possible onset of cavitation in tissue.

Peak negative pressure

It is generally accepted that the peak pressure in the ultrasonic wave is an important determinant of the potential of the wave to initiate cavitation. At the high acoustic output generated by modern equipment the ultrasonic wave propagating in water undergoes a progressive distortion. The wave becomes asymmetric, the positive component developing a steep shock front while the negative component incurs a stretching form of distortion. In a fully developed shockwave the peak positive pressure may be three times greater than the peak negative pressure. Accurate measurement of the fast rising shock wavefront, however, is difficult and requires specialized calibration equipment.

The steep slope of the shock wave indicates that the wave contains beside the fundamental also many higher harmonic frequencies. It is not clear to what extent the development of shock waves occurs in soft tissue. These preferentially attenuate the higher frequencies, negating the development of the shock wave. Because of the difficulty in measuring the amplitude of the positive peak pressure accurately and the uncertainty in the degree of development of the shock wave in tissue, the current practice is to use the peak negative pressure as the descriptor of the peak pressure in the wave.

2.2.2.2.6 Mechanisms of interaction

2.2.2.2.6.1 Introduction

During clinical ultrasonographic examinations acoustic energy is transmitted into the body and interacts with its tissues in ways that may result in a measurable biological response. The best understood mechanism of interaction is that involving heating. Some of the ultrasound energy is reflected back from interfaces between biological tissues to produce the echographic images, while some of the energy is absorbed and converted to heat. The amount of heat generated is mostly dependent upon the ability of the tissue to absorb, rather than reflect or disperse, ultrasonic energy (i.e. its absorption coefficient) and this is a function of its gross molecular structure. Generally, more dense materials such as bone and teeth have high acoustic absorption coefficients and are heated to a greater extent than soft tissue.

Cavitation is a non-thermal mechanism of interaction that involves the formation, oscillation and occasional collapse of bubbles in a sound field. The subject is comprehensively described by Leighton [94Lei]. Another non-thermal effect is acoustic streaming.

2.2.2.2.6.2 Thermal mechanism

A large amount of data has been published on the estimated temperature elevation in tissue resulting from exposure by diagnostic ultrasound equipment. One theoretical model developed by the National Council for Radiation Protection and Measurement (NCRP) estimated the worst-case steady state heating that would never be exceeded in diagnostic ultrasound examinations [92NCR]. Another model proposed by the American Institute of Ultrasound in Medicine and the National Electrical Manufacturers Association (AIUM/NEMA) estimated values that are not expected to be exceeded in the majority of ultrasonographic examinations [92AIU]. The AIUM/NEMA model provides estimates known as the thermal index (TI) that may be displayed during equipment operation. This provides the operator with a relative value from which assumptions may be made about the risk of producing a thermally related bioeffect during an ultrasonographic examination.

A number of studies have been carried out to quantify ultrasound-induced temperature increase in tissue under simplified, stable experimental conditions where there is no blood flow. These studies use (1) specimens of dead soft tissue exposed in test tanks; (2) specimens of bone exposed in test tanks; and (3) models of gel materials and bone that mimic biological tissue [92Bar1, 93Bos].

Concerning thresholds for biological effects resulting from thermal mechanisms it is referred to Sect. 2.3.2.2

2.2.2.2.6.3 Non-thermal mechanisms

Cavitation

Cavitation refers to the formation and dynamic behaviour of gas bubbles. The term ‘cavitation’ was first mentioned in the hydrodynamic context of the observation of separation of water from a ship’s propeller in motion. Acoustic cavitation is the growth of bubbles from microscopic pockets of undissolved gas and/or vapour, stabilized on solid surfaces or by surface-active films, due to the reduction in pressure in the negative phase of the acoustic cycle.

The inception of acoustic cavitation is demarcated by a specific threshold value: the minimum acoustic pressure necessary to initiate the growth of a cavity in a fluid during the rarefaction phase of the cycle. A number of parameters affect this threshold, including: initial bubble or ‘nucleus’ size, acoustic pulse characteristics (such as centre frequency, pulse repetition frequency and pulse duration), ambient hydrostatic pressure and host fluid properties (such as density, viscosity, compressibility, heat conductivity and surface tension). Inertial, formerly called ‘transient’, cavitation is the growth of short-lived bubbles which undergo large variations from their equilibrium sizes during pressure changes in a few acoustic cycles. During contraction, the surrounding fluid inertia controls the bubble motion. Large acoustic pressures are necessary to generate inertial cavitation, and the collapse of these cavities is often violent.

At high-peak negative pressures inertial cavitation can cause cavities or bubbles in liquids to collapse and release energy sufficiently intense to disrupt molecular bonds and produce chemically reactive free radicals [83NCR]. These free radicals may interfere with DNA, causing chromosomal damage. Although this has been reported in cell studies, it has not been directly observed in patients or in laboratory animals exposed to diagnostic intensities.

It is possible to generate bubbles in vivo from the short-duration, high-amplitude acoustic pulses used by extracorporeal shock wave lithotripters (ESWL). The peak positive or compressional pressure (p_c) for lithotripsy pulses can be as much as 50 MPa and the negative, rarefactional pressure (p_r) around 20 MPa. Although ESWL pulses have significant energy at high frequencies due to finite amplitude distortion, a large portion of the energy is actually in the 100-kHz frequency range, much lower than that of diagnostic ultrasound scanners. The lower frequency makes cavitation more likely. There is evidence to indicate that the destruction of gall stones or renal calculi may be due to the effects of collapsing bubbles [89Wil]. As evidence of the destructive power, when cavitation due to application of extracorporeal shock wave lithotripsy occurs next to a solid surface, a high-velocity liquid jet may form during collapse, generating a sufficient impulse to produce observable damage on metal surfaces [88Cru]. If one takes a sheet of aluminium foil and places it at the focus of a lithotripter, small pinholes will be generated.

Lithotripsy and diagnostic ultrasound differ in the acoustic power generated and are not comparable in the bioeffects produced. Some diagnostic devices produce peak rarefactional pressures greater than 3 MPa, which is in the lower range of lithotripter output [94Pat].

Acoustic streaming

Modern diagnostic ultrasound equipment produces sufficient radiation pressure to cause bulk movement of fluids along the ultrasound propagation path (acoustic streaming). Low-level radiation stresses (forces) are always exerted on the tissues by the pulses of ultrasound used in ultrasonic scanning. These forces are sufficient to cause fluid to flow away from the transducer along the path of the ultrasound beam, within volumes of at least 1 ml [89Sta]. There is no evidence of hazard associated with this streaming when generated within amniotic fluid or urine. Within soft tissues the radiation stress is probably insufficient to cause gross damage at diagnostic intensities, although the effect of stress enhancement near large acoustic boundaries requires further study. Radiation pressure from ultrasonic pulses of diagnostic amplitude, but longer duration, has been shown to cause compression ischemia, and can be sensed by the skin. Weakly tethered cell structures, and in particular embryonic tissue, could bear the risk of disturbance if the recovery from the application of a transient stress wave were incomplete. Any decision to use diagnostic ultrasound during the first trimester should acknowledge radiation stress as a potential bioeffects mechanism and, prudently, reduced exposure levels should be used.

2.2.2.2.7 Summary

Different acoustic parameters are used to describe ultrasound exposure. Some of these parameters can be used to quantify the risk of biological effects that may occur during ultrasound diagnostic.

There is a large variety of medical ultrasound specialties including cardiology, vascular disease, ophthalmology, internal medicine gynaecology and obstetrics. The predominant medical diagnostic application of ultrasound is in the frequency range 2-10 MHz, with occasional extension to 20 MHz for some eye examinations. Ultrasound irradiation falls into two categories – pulsed fields and continuous wave – the latter being used for Doppler frequency-shift identification of moving structures and for blood flow velocity measurement.

There are thermal and non-thermal interaction mechanisms. During clinical ultrasonographic examinations acoustic energy is transmitted into the body and interacts with its tissues in ways that may result in a measurable biological response. The best understood mechanism of interaction is that involving heating. Cavitation is a non-thermal mechanism of interaction that involves the formation, oscillation and occasional collapse of bubbles in a sound field. Ultrasound contrast agents can increase the likelihood of occurrence and the extent of ultrasound induced bioeffects *in vivo*. Modern diagnostic ultrasound equipment produces sufficient radiation pressure to cause bulk movement of fluids along the ultrasound propagation path (acoustic streaming).

2.2.2.3 Static and slowly varying electric and magnetic fields

2.2.2.3.1 Physical characteristics

When a voltage is applied to an object such as an electrical conductor, the conductor becomes charged and forces start to act on other charges in the vicinity. Two types of forces may be distinguished: that which arises from stationary electric charges, known as the electrostatic force, and that which appears only when charges are moving (as in an electric current in a conductor), known as the magnetic force. The concept of field has been created to describe the existence and spatial distribution of these forces. Reference is then made to field of force, or simply electric and magnetic fields.

The term static refers to a situation where all charges are fixed in space, or move as a steady flow, so that both charges and current densities are constant in time. For fixed charges there is an electric field whose strength at any point in space depends on the value and geometry of all both the electric and the

charges. For a steady current in a circuit, magnetic field is constant in time (static fields), since the charge density at any point of the circuit is constant.

Static electric and magnetic fields are characterized by steady, time independent strengths and correspond to the zero-frequency limit of the extremely low frequency (ELF) band. Electricity and magnetism are distinct phenomena as long as charges and current are static.

Quantities and Units

A magnetic field refers to the fields of force, produced by moving electric charges (electric currents), that act on other moving charges. The field from a permanent magnet results from the subatomic spin of electrons. A magnetic field is a vector field, and the fundamental vector quantities describing the magnetic field are the field strength H and the magnetic flux density B (or equivalently, the magnetic induction). The magnetic flux density is related to the magnetic field strength by the formula $H = B/\mu$. The value of μ (the magnetic permeability) is determined by the properties of the medium. In biological material, the magnetic permeability is equal to μ_0 , the value of the permeability of free space (air). Thus, the values of B and H for biological materials are related by this constant.

An electric field refers to a region near an electric charge in which a force is exerted on a charged particle. The force between two point charges is described by Coulomb's law. The electric field is denoted by E and is a vector quantity. The SI unit for E is Newton per coulomb [$N C^{-1}$].

However, it is easier to measure the electric potential V rather than the force and charge, and the unit of volt per meter ($V m^{-1}$) is used in practice. As electric fields exert forces on charged particles, this will cause an electric current to flow in an electrically conductive material. This current is specified by the current density J , with a unit of ampere per square meter [$A m^{-2}$].

The quantities, units, and symbols used in describing electric and magnetic fields are provided in [Table 2.2.2.3](#).

Table 2.2.2.3. Electric and magnetic field quantities and units in the SI system

Quantity	Symbol	Unit
Electric field strength	E	volt per meter [$V m^{-1}$]
Electric flux density	D	Coulomb per square meter [$C m^{-2}$]
Current	I	ampere [A]
Current density	J	ampere per square meter [$A m^{-2}$]
Magnetic field strength	H	ampere per meter [$A m^{-1}$]
Magnetic flux	φ	weber [Wb] = [$V s$]
Magnetic flux density	B	tesla ^{a)} [T] = [$Wb m^{-2}$]
Permeability	μ	henry per meter [$H m^{-1}$]
Permeability of vacuum	μ_0	$\mu_0 = 1.257 \times 10^{-6} H m^{-1}$

^{a)} 1 T = 10^4 gauss (G), a unit in the CGS unit system

2.2.2.3.2 Natural and anthropogenic sources

Static electric fields occur naturally in the atmosphere. Values of up to $3 kV m^{-1}$ can occur under thunderclouds, but otherwise lie in the range $1-100 V m^{-1}$ in fair weather. Direct current (DC) power transmission can produce static electric fields of up to $20 kV m^{-1}$, rail systems using DC can generate fields of up to $300 V m^{-1}$ inside the train, and video display units (VDUs) create electric fields of around $10-20 kV m^{-1}$ at a distance of 30 cm.

The geomagnetic field varies over the Earth's surface between about $30-70 \mu T$. Magnetic flux densities of up to $2 mT$ have been reported inside electric trains and in developmental magnetic levitation (MagLev) transport systems. Workers in the electrolytic reduction of alumina can be locally exposed to larger fields of up to around $60 mT$, and DC electric arc welding produces around $5 mT$ at 1 cm from the

welding cables. Strong fields are also produced in high-energy technologies such as thermonuclear reactors, magneto-hydrodynamic systems and superconducting generators. Research facilities that use bubble chambers, particle accelerators, superconducting spectrometers, and isotope separation units also have areas with high magnetic flux densities around these devices. Workers can be locally exposed regularly and for long periods to fields as high as 1.5 T. The advent of superconductors in the 1970s and 1980s also facilitated the use of much larger magnetic fields in medical diagnosis through the development of magnetic resonance imaging (MRI) and spectroscopy. The static magnetic field of MRI scanners in routine clinical systems is in the range of 0.2-3 T [05Gow]. Functional MRI using magnetic fields up to about 10 T is now widely used in academic and medical research on human brain function. Occupational exposure up to and exceeding 1 T can occur during the construction and testing of these devices and during medical procedures carried out in interventional MRI.

2.2.2.3.3 Interaction mechanisms

Static electric fields

Exposure to a static electric field does not induce an internal field in the body because of the high conductivity of body tissues relative to air. The main action of a static electric field is the generation of a surface charge on the exposed object. Such charge build-up can also be generated by friction, e.g., when walking on a carpet. This charge build-up is more effective in dry air because of its lower conductivity. These charges may result in local electric fields strengths of up to about 500 kV m^{-1} . The effects that result from the build-up of surface charge are movement of hairs and spark discharges. Spark discharges will occur when a person who is well insulated from the ground touches a grounded object or when a grounded person touches a conductive object that is well insulated from ground. However, the threshold static electric field values will vary depending on the degree of insulation and other factors.

Static magnetic fields

The following three classes of physical interactions of static magnetic fields with biological systems are well established on the basis of experimental data: electrodynamical effects, magnetomechanical effects, and effects on radical pair formation [09ICN2].

Electrodynamical effects

When charges move through a static magnetic field, electric flow potentials and electric currents are generated [05Sch]. This will occur when a body moves through a static magnetic field but also occurs in a stationary body as a result of the flow of blood through the field. These flow potentials and currents can result in a magnetohydrodynamic force, which may slow the flow of blood. The effect will be most prominent in the larger blood vessels, since the larger the diameter of the vessel, the more charged particles flow through and consequently the higher the flow potential and induced current. In the aorta, magnetomechanical effects have been calculated to result in a 5% decrease in blood flow with exposure to a 10 T field and 10% in a 15 T field [96Kin].

Other effects of flow potential may occur in the heart. Theoretical calculations [05Hol] raise three possibilities: minor changes in heartbeat (which may be considered to have no health consequences), the induction of ectopic heartbeats (which may be more physiologically significant), and an increase in the likelihood of re-entrant arrhythmia (possibly leading to ventricular fibrillation). The first two effects are thought to have thresholds in excess of 8 T, and threshold values for the third are difficult to assess at present because of modelling complexity. Some 5-10 per 10.000 people are particularly susceptible to re-entrant arrhythmia, and the risk to such people may be increased by exposure to strong static magnetic fields and pulsed gradient fields.

Movement of the whole or part of the body, eyes and head, in a static magnetic field gradient will also induce an electric field and current during the period of movement. Dosimetric calculation suggests that such induced electric fields will be substantial during normal movement around or within fields $> 2\text{-}3 \text{ T}$, and may account for the numerous anecdotal reports of vertigo and nausea and occasionally magnetic phosphenes experienced by patients, volunteers, and workers moving in the field. Calculations have been performed for a body moving near and inside an MRI magnet [05Cro]. For example, for a body moving at

a constant speed of 0.5 m s^{-1} in a 4 T magnet, the maximum induced electric field strength is approximately 2 V m^{-1} and the induced current density is approximately 300 mA m^{-2} . However, the magnitude and frequency dependence of the induced fields that result during normal body movement in the strong static magnetic field gradients associated with MRI systems above 2-4 T have not been quantified.

Magnetomechanical effects

These effects can be distinguished in alignment of polarized structures in the magnetic field and movement of ferrometallic objects.

The first effect may lead to orientation of magnetically anisotropic structures in uniform fields. Since the amount of magnetically anisotropic material in the human body is minimal, this effect is not considered of importance for human health.

Movement of ferrometallic objects may cause problems, however. First as a result of forces and torques exerted on metallic objects carried in or on the body, such as pacemakers and metallic implants. People fitted with such devices should be kept at sufficient distance from strong magnetic fields. A second effect of this sort is the translation of paramagnetic and ferromagnetic materials in magnetic field gradients. There are many examples of small and large metallic objects that were propelled into the bore of an MRI machine. Clearly, this is an important safety issue.

Radical pair mechanism

Since radicals are considered to be effective in damaging biological molecules, an effect on the lifetime of radicals might ultimately have an effect on health. It is known that spin-correlated radical pairs in the singlet state will recombine to form reaction products, unless they convert to the triplet state. This singlet-triplet interconversion may be affected by static magnetic fields. For the interaction to be of any significance, two conditions should be met: first the radicals should obviously recombine, and second the recombination period should be sufficiently long ($> 100 \text{ ns}$).

Although there remain biological systems worthy of further investigation, there is, based on the evidence at present, no strong likelihood of major effects of physiological consequence on cellular function or of long-term mutagenic effects arising from magnetic-field induced changes in free radical concentrations or fluxes [05Hor].

2.2.2.3.4 Summary

Static electric fields occur naturally in the atmosphere. Values of up to 3 kV m^{-1} can occur under thunderclouds, but otherwise lie in the range $1\text{-}100 \text{ V m}^{-1}$ in fair weather. Direct current (DC) power transmission can produce static electric fields of up to 20 kV m^{-1} , rail systems using DC can generate fields of up to 300 Vm^{-1} inside the train, and video display units (VDUs) create electric fields of around $10\text{-}20 \text{ kV m}^{-1}$ at a distance of 30 cm.

The geomagnetic field varies over the Earth's surface between about $35\text{-}70 \mu\text{T}$. Magnetic flux densities of up to 2 mT have been reported inside electric trains. Workers in the electrolytic reduction of alumina can be locally exposed to larger fields of up to around 60 mT. The advent of superconductors in the 1970's and 1980's facilitated the use of much larger magnetic fields in medical diagnosis through the development of magnetic resonance imaging (MRI) and spectroscopy. The static magnetic field of MRI scanners in routine clinical systems is in the range of 0.2-3 T. In research applications, magnetic fields up to 8 T are used for whole body patient scanning. Occupational exposure up to and exceeding 1 T can occur during the construction and testing of these devices and during medical procedures carried out in interventional MRI.

The following three classes of physical interactions of static magnetic fields with biological systems are well established on the basis of experimental data:

- electrodynamic interactions with ionic conduction currents;
- magnetomechanical effects, including the orientation of magnetically anisotropic structures in uniform fields and the translation of paramagnetic and ferromagnetic materials in magnetic field gradients; and
- effects on electronic spin states of reaction intermediates.

Forces and torques on both endogenous and exogenous metallic objects are the interaction mechanism of most concern. The induction of electric fields and currents in tissue is also of concern. Other interaction mechanisms do not appear to be of concern at this stage. None of the mechanisms discussed to date would seem to indicate differences in effects from acute or chronic exposure, although no suitable epidemiological studies are available in this regard.

2.2.2.4 Time-varying electric and magnetic fields of frequencies less than 100 kHz

2.2.2.4.1 Introduction

The frequencies under consideration range from above 0 Hz to 100 kHz. By far the majority of studies have been conducted on power-frequency (50 or 60 Hz) magnetic fields, with a few studies using power-frequency electric fields. In addition, there have been a number of studies concerning very low frequency (VLF, 3-30 kHz) fields, switched gradient magnetic fields used in magnetic resonance imaging, and the weaker VLF fields emitted by visual display units and televisions.

Electric and magnetic fields exist wherever electricity is generated, transmitted or distributed in power lines or cables, or used in electrical appliances. Since the use of electricity is an integral part of our modern lifestyle, these fields are ubiquitous in our environment.

The unit of electric field strength is volts per meter [$V\ m^{-1}$] or kilovolts per meter [$kV\ m^{-1}$] and for magnetic fields the flux density is measured in tesla [T], or more commonly in millitesla [mT] or microtesla [μT] is used.

2.2.2.4.2 Natural and anthropogenic sources

Naturally occurring fields

The natural electric field encountered above the surface of the Earth varies greatly with time and location. The primary cause of the field is the charge separation that occurs between the Earth and the ionosphere, which acts as a perfect conductor separated by air of negligible conductivity [81Kön]. The field near the surface in fair weather has a typical strength of about $130\ V\ m^{-1}$. The strength generally depends on height, local temperature, humidity profile and the presence of ions in the atmosphere. Variations of up to $40\ kV\ m^{-1}$ occur near thunderstorms, although even in the absence of local lightning, fields can reach up to $3\ kV\ m^{-1}$.

The Earth's magnetic field changes continually at periods ranging from a few milliseconds up to 10^{12} seconds. The main feature of the geomagnetic field is its close resemblance to a dipole field aligned approximately with the spin axis of the Earth. The dipole field is explained by electrical currents that flow in the core. The vertical component of the field reaches a maximum of about $70\ \mu T$ at the magnetic poles, and approaches zero at the magnetic equator, conversely the horizontal component is close to zero at the poles and has a maximum just over $30\ \mu T$ at the magnetic equator.

The main part of the Earth's magnetic field across the ELF and VLF part of the spectrum have their primary cause outside the Earth, associated with processes in the ionosphere and magnetosphere (Garland, 1979). These include the regular solar and lunar daily variations upon which more irregular disturbances are superimposed.

The ELF variations arise mainly from the effects of solar activity in the ionosphere and atmospheric effects such as lightning discharges which cause resonance oscillations in the Earth-ionosphere cavity. Changes in ELF signals over 11-year and 27-day periods and circadian variations reflect the solar influences [96EC]. The electromagnetic fields that arise from lightning discharges, commonly known as

atmospherics, have a very broad frequency range with spectral components from below 1 Hz up to a few megahertz.

Artificial fields

Electric and magnetic fields exist wherever electricity is generated, transmitted or distributed in power lines or cables, or used in electrical appliances. Since the use of electricity is an integral part of our modern lifestyle, these fields are ubiquitous in our environments.

Residential exposure to power frequency magnetic fields does not vary dramatically across the world. The geometric mean magnetic field in homes ranges between 0.025 and 0.07 µT in Europe and 0.055 and 0.11 µT in the USA. The mean values of electric field in the home are in the range of several tens of volts per meter. In the vicinity of certain appliances, the instantaneous magnetic-field values can be as much as a few hundred microtesla. Near power lines, magnetic fields reach approximately 20 µT and electric fields up to several thousand volts per meter.

Occupational exposure, although predominantly to power-frequency fields, may also include contributions from other frequencies. The average magnetic field exposures in the workplace have been found to be higher in “electrical occupations” than in other occupations such as office work, ranging from 0.4-0.6 µT for electricians and electrical engineers to approximately 1.0 µT for power line workers, with the highest exposures for welders, railway engine drivers and sewing machine operators (above 3 µT). The maximum magnetic field exposures in the workplace can reach up to approximately 10 mT in the vicinity of induction furnaces and equipment heaters and this is invariably associated with the presence of conductors carrying high currents. In the electrical supply industry, workers may be exposed to electric fields up to 30 kV m⁻¹.

The main issue in the ELF range regarding MRI exposures to workers, including clinicians and maintenance engineers, are induced fields and currents due to the switched gradient coils (induced body field strengths and currents, see below). It has been estimated that workers may be exposed to gradient fields up to 10 mT, at an equivalent frequency of around 500 Hz [05Hil]. This value is 200 times the reference level in the ICNIRP Guideline [98ICN].

2.2.2.4.3 Interaction mechanisms

Direct and indirect mechanisms of interaction

Exposure of the body to externally applied ELF electric and magnetic fields results in the production of fields internal to the body. Mechanisms that describe such interactions include:

- Electric fields act on an exposed human or animal, resulting in the application of (often intensified) electric fields to the outer surface of the body of the subject, and the induction inside the body of electric fields and currents.
- Magnetic fields act on an exposed human or animal, resulting in magnetic field penetration throughout the body, the induction of electric fields and currents inside the body, and the application of forces to moving charges within the body.
- Electric currents induced in a conducting object (e.g. an automobile) exposed to an electric field can pass to and through a human or animal in contact with it.
- Magnetic field coupling to a fence line or other conductors can cause currents to pass through a human or animal in contact with.
- A human or animal that is standing on earth that is carrying electric currents may be exposed to a “step potential” that will cause currents to flow in the body.
- Transient (often called spark) discharges can occur when two bodies exposed to a strong electric field come into very close proximity and/or at the instant of their contact.

It is useful to classify coupling mechanisms between humans or animals and electric and magnetic fields into indirect or direct forms of coupling, depending on whether the presence of a second body, in addition to that of exposed organism, is required for the coupling to occur. By this definition the first two

mechanisms listed above are examples of direct coupling between living organisms and fields because they can occur when only the exposed organism is present.

The latter four mechanisms listed above are examples of indirect coupling mechanisms since they can occur only when the exposed organism is in the vicinity of other bodies. These bodies can include the ground, other humans or animals, and/or objects such as automobiles, fences, etc.

The following sections discuss the direct coupling of electric and magnetic fields with living organisms in more detail. For a discussion of indirect interactions, including contact currents and electric shocks, the reader is referred to [87Ten, 88Ber].

Direct electric field coupling

As in the case of static electric fields, time-varying electric fields exert forces on charged particles and therefore induce an electric surface charge in an electrically conductive material such as living tissue. As the field oscillates, the induced surface charge density will oscillate correspondingly, thus requiring the presence of currents inside the object. Thus, both currents and electric fields are induced inside a conducting object by an external electric field.

In the ELF range, the variation in surface charge density is so slow that the currents and fields generated inside the objects are very small. Estimates show that the ratios of fields inside the body to those applied externally are ca. 10^{-12} at DC, 10^{-6} to 10^{-8} at 50 Hz and 10^{-3} to 10^{-4} at 100 kHz. ELF fields exceeding approximately 20 kV m^{-1} can be perceived by nearly all people as a result of surface charge induced on the body. However, the threshold field level for detection can be lowered significantly (to about 10 kV m^{-1}) by extending the finger tips in the direction of the electric field source, thereby causing field enhancement in the region of the fingers, hand and forearm. In high-voltage ELF fields, the body hair oscillates at a frequency that is equal to twice the frequency of the applied field. In addition, ELF field strengths greater than approximately $5 - 7 \text{ kV m}^{-1}$ can produce a variety of effects such as shock reactions associated with spark discharges and contact currents from closely approaching or touching ungrounded conductors within the field.

Useful information can be obtained from experiments with conducting models that simulate the shapes of humans and animals. In recent years, various computational methods have been used to evaluate induced electric fields in high-resolution models. Computations of exposure to electric fields are generally more difficult than for the magnetic field exposure, since the human body significantly perturbs the exposure field. The methods that have been successfully used so far for high-resolution dosimetry are the finite difference (FD) method in frequency domain and time domain (FDTD) and the finite element method (FEM). Each method and its implementation offer some advantages and have limitations, as reviewed by [00Stu]. Some of the methods and computer codes have undergone extensive verification by comparison with analytic solutions [96Daw]. An extensive valuation of accuracy of various dosimetric measures is also available [01Daw].

Several numerical computations of the electric field and current density induced in various organs and tissues have been performed [00Dim, 01Hir, 05Dim]. Organs small in any dimension are poorly-represented by large voxels. The maximum induced quantities are consistently higher as the voxel dimension decreases.

The main features of dosimetry for exposures to the ELF electric fields can be summarized as follows:

- Magnitudes of the induced electric fields are typically 10^{-4} to 10^{-7} of the magnitude of external unperturbed field.
- Since the exposure is mostly to the vertical field, the predominant direction of the induced fields is also vertical.
- In the same exposure field, the strongest induced fields are for the human body in contact through the feet with a perfectly conducting ground plane, and the weakest induced fields are for the body in free space, i.e., infinitely far from the ground plane.
- The global dosimetric measure of short-circuit current for a body in contact with perfect ground is determined by the body size and shape (including posture) rather than tissue conductivity.
- The induced electric field values are to a lesser degree influenced by the conductivity of various organs and tissues than are the values of the induced current density.

Direct magnetic field coupling

Magnetic fields interact with living systems by two well established mechanisms: (1) induced electric fields produced in accord with Faraday's law of magnetic induction, and (2) direct magnetic field effects on magnetic particles such as the crystals of magnetite (Fe_3O_4) that have been found in a number of organisms. A large number of other possible interaction mechanisms, including resonance effects, have been proposed. However, the evidence for the existence of these interactions in living systems is not well established. The physical basis and biological implications of the first mechanism mentioned above are discussed in the following sections of this chapter.

Magnetically induced electric fields

ELF magnetic fields induce electrical currents in tissue that circulate in loops within planes that are orthogonal to the direction of incidence of the field. This relationship between a time-varying magnetic field and the circulating electric field that it induces is expressed formally by Faraday's law:

$$\partial\mathbf{B}/\partial t = - \operatorname{rot} \mathbf{E} \quad (2.2.2.7)$$

where $\operatorname{rot} \mathbf{E}$ is the curl of the electric field vector. A magnetically induced electric field gives rise to currents that are predicted from Ohm's law, $\mathbf{J} = \sigma \mathbf{E}$, where \mathbf{J} is the induced current density vector, expressed in A m^{-2} , and σ is the tissue electrical conductivity in S m^{-1} .

For the specific case of a circular loop of tissue with radius R [m] intersected by a spatially uniform, time-varying magnetic field orthogonal to the loop, Faraday's law gives the magnitude of the peak electric field tangent to the loop as:

$$E_{\text{peak}} = (R/2) dB/dt \quad (2.2.2.8)$$

If the magnetic field is sinusoidal with amplitude B_0 and frequency f , then $B = B_0 \sin(2\pi ft)$ and from eq. (2.2.2.8) and Ohm's law, the peak current density induced in tissue with an average conductivity σ is given by

$$J_{\text{peak}} = \pi f R B_0 \sigma \quad (2.2.2.9)$$

The application of this relationship between the induced current density and the applied magnetic field given in eq. (2.2.2.9) is the basis for modern computational dosimetry techniques. Similarly, ellipsoidal loops can be considered to better fit into the body shape.

Electric fields and currents induced in the human body cannot be measured easily. Measurements in animals have been performed, but data are limited, and the accuracy of measurements is relatively poor.

Heterogeneous models of the human body similar to those used for electric field exposures have been numerically analyzed using the impedance method (IM) [01Gan], and the scalar potential finite difference (SPFD) technique [96Daw, 98Dim]. Even more extensive data than for the electric field are available for the magnetic field. The influence on the induced quantities of the model resolution, tissue properties in general and muscle anisotropy specifically, field orientation with respect to the body, and to a certain extent body anatomy have been investigated [98Dim, 00Stu, 05Dim]. In the past, the maximum current density in a body part has often been calculated using the largest loop of current that can be incorporated in it. Dawson et al. [99Daw] have shown that induced parameters should be calculated for organs *in situ* instead of for isolated ones, since there is a significant influence of surrounding structures.

The main features of dosimetry for exposures to the uniform ELF magnetic field can be summarized as follows.

- The electric fields induced in the body depend on the orientation of the magnetic field with respect to the body.
- For most organs and tissues, as expected, the magnetic field orientation normal to the torso (front-to-back) gives maximum induced quantities.
- In the brain, cerebrospinal fluid, blood, heart, bladder, eyes and spinal cord, the highest quantities are induced by the magnetic field oriented side-to-side.
- Consistently lowest induced fields are for the magnetic field oriented along the vertical body axis.

- For a given field strength and orientation, greater electric fields are induced in a body of a larger size.
- The induced electric field values are to the lesser degree influenced by the conductivity of various organs and tissues than the values of the induced current density.

Although an initial physical mechanism of interaction of time-varying magnetic fields with living tissues is the induction of internal electric fields and currents, the triggering of biological effects by these electrical signals must involve subsequent electrochemical processes that influence cellular functional properties. There is now a relatively strong body of evidence suggesting that the transduction processes through which induced electric signals influence cellular properties involve interactions at the level of the cell membrane [03ICN]. It has been demonstrated, for example, that a magnetically-induced electric field of 0.1 V m^{-1} in the extracellular fluids can significantly increase Ca^{2+} uptake in mitogen-activated thymocytes [90Wall]. A number of other membrane effects and cellular manifestations of ELF field interactions with cell membranes, many of which occur at threshold levels of induced fields below 0.1 V m^{-1} are discussed in [03ICN, 07WHO].

As mentioned above, the main issue in the ELF range regarding MRI exposures to workers, including clinicians and maintenance engineers, are induced fields and currents due to the switched gradient coils. Several authors have examined the question of induced fields and currents in workers standing at or near the end of the bore of MRI units. In particular, Crozier et al. used the voxel models to calculate field strengths and current density values in specific tissues at particular distances from the ends of the gradient coils [07Cro]. An electric field strength of 700 mV m^{-1} in the brain due to exposure to the gradient field was reported. An adjustment for more realistic rise time (0.25 ms) in practical situations resulted in approximately 500 mV m^{-1} . Although the ICNIRP basic restriction would be exceeded, there would still be a margin of 4 between the brain electric field and the effects threshold (see also Sect. 5.4.2.1). There would thus be some justification for allowing this safety margin to be exceeded under brief but specified circumstances where patient well-being is concerned (see Sect. 5.4).

2.2.2.4.4 Summary

Electric and magnetic fields exist wherever electricity is generated, transmitted or distributed in power lines or cables, or used in electrical appliances. Residential exposure to power-frequency magnetic fields does not vary dramatically across the world. The geometric-mean magnetic field in homes ranges between 0.025 and $0.07\text{ }\mu\text{T}$ in Europe and 0.055 and $0.11\text{ }\mu\text{T}$ in the USA. In the vicinity of certain appliances, the instantaneous magnetic-field values can be as much as a few hundred microtesla. Near power lines, magnetic fields reach approximately $20\text{ }\mu\text{T}$ and electric fields up to several thousand volts per meter. The mean values of the electric field in the home are in the range of several tens of volts per meter.

Occupational exposure, although predominantly to power-frequency fields, may also include contributions from other frequencies. The average magnetic field exposures in the workplace have been found to be higher in “electrical occupations” than in other occupations such as office work, ranging from 0.4 - $0.6\text{ }\mu\text{T}$ for electricians and electrical engineers to approximately $1.0\text{ }\mu\text{T}$ for power line workers, with the highest exposures for welders, railway engine drivers and sewing machine operators (above $3\text{ }\mu\text{T}$). The maximum magnetic field exposures in the workplace can reach approximately 10 mT and this is invariably associated with the presence of conductors carrying high currents. In the electrical supply industry, workers may be exposed to electric fields up to 30 kV m^{-1} .

The main issue in the ELF range regarding MRI exposures to workers, including clinicians and maintenance engineers, are induced fields and currents due to the switched gradient coils. It has been estimated that workers may be exposed to gradient fields up to 10 mT , at an equivalent frequency of around 500 Hz.

Exposure to external electric and magnetic fields at extremely low frequencies induces electric fields and currents inside the body. Electric fields induced in tissue will directly stimulate single myelinated nerve fibres in a biophysically plausible manner when the internal field strength exceeds a few volts per meter. Much weaker fields can affect synaptic transmission in neural networks as opposed to single cells.

Such signal processing by nervous systems is commonly used by multicellular organisms to detect weak environmental signals. A lower bound on neural network discrimination of 1 mV m^{-1} has been suggested, but based on current evidence, threshold values around $10\text{-}100 \text{ mV m}^{-1}$ seem to be more likely.

With regard to indirect effects, the surface electric charge induced by electric fields can be perceived, and it can result in painful microshocks when touching a conductive object.

2.2.2.5 Electromagnetic fields of frequencies above 100 kHz

2.2.2.5.1 Physical characteristics

The physical properties of radiofrequency radiation (RFR) are treated in detail in [81NCR, 03AGN, 09ICN2]. RFR is defined by the International Telecommunication Union (ITU) as being part of the electromagnetic spectrum used for wireless communication, including frequencies that range between 3 Hz and 300 GHz. The 300 MHz-300 GHz frequency range is also known as the microwave range. This sections deal with the frequency range between 100 kHz and 300 GHz. High-frequency radiation is emitted as continuous single-frequency radiation or amplitude-, frequency- or pulse-modulated radiation.

High frequency electromagnetic fields are quantified in terms of the electric field strength \mathbf{E} , expressed as volts per meter [$\text{V}\cdot\text{m}^{-1}$] and magnetic field strengths \mathbf{H} , expressed as amperes per meter [$\text{A}\cdot\text{m}^{-1}$]. \mathbf{E} and \mathbf{H} are vector fields. In the far field of an antenna, the high frequency electromagnetic field is often quantified in terms of power flux density S , expressed in units of watt per square meter [$\text{W}\cdot\text{m}^{-2}$].

The oscillation energy transported from the source of radiation, e.g. the antenna, is largely described as the power density, S [$\text{W}\cdot\text{m}^{-2}$]. With pulsed radiation, the peak power flux densities are likely to be 100 times stronger than power densities of a continuous radiation of similar mean power density. When characterising the electromagnetic field, near- and far-field zones must be taken into account. The start of the far-field radiation zone can be determined approximately by the relation $r = D^2/\lambda$ (where r is the distance from the antenna, D is the largest antenna diameter, and λ is the wavelength). In the far-field zone, the power density decreases by $1/r^2$; only here the relation is $S = E^2/377 = 377 H^2$, i.e., it is enough to know a single parameter S , E or H to accurately define the irradiation situation. In the near field range, E as well as H must be known, together with their appropriate phase differences, in order to identify precisely the irradiation situation.

For the purpose of radiation protection, physical quantities to describe sources and field properties as well as the interaction of such fields with biological systems are needed, to quantify the exposure of the human body to non-ionizing radiation and to estimate the absorbed energy and its distribution inside the body (dosimetric quantities)

A dosimetric measure that has been widely adopted is the specific absorption rate (SAR), defined as the time derivative of the incremental energy δW , absorbed by or dissipated in an incremental mass, δm , contained in a volume element, δV , of a given density ρ :

$$\text{SAR} = \delta/\delta t (\delta W/\delta m) = \delta/\delta t (\delta W/\rho\delta V) \quad (2.2.2.10)$$

The SAR is expressed in watt per kilogram (W kg^{-1}). SAR can also be averaged over smaller masses (e.g., organ, eye, 1 or 10 g of tissue). A value of the whole body SAR of about 1 W kg^{-1} corresponds with the average basic metabolic rate in man.

Exposure metrics and dosimetric quantities

At the physical level, RF exposure may be specified in terms of source type, continuous-wave frequency, modulation, zone of exposure (near or far field), partial- or whole-body exposure, incident electric field and magnetic field strengths, incident power density (when appropriate), peak or averaged value, and duration of exposure.

The coupling of RF electromagnetic energy into biological systems may be quantified by the induced electric and magnetic fields, power deposition, energy absorption, and their distribution and penetration into biological tissues. These quantities are all functions of the source and its frequency or wavelength and their relationship to the physical configuration and dimension of the biological body.

The metrics of specific absorption rate (SAR) and specific absorption (SA) in biological systems or tissue models have been adopted both by ICNIRP [98ICN] and IEEE [96IEE] as the dosimetric quantities. The metric SAR in W kg^{-1} is defined in eq. (2.2.2.10). Specific absorption in J kg^{-1} is the total amount of energy deposited or absorbed and is given by the integral of SAR over a finite interval of time. Knowledge of induced fields, SA, and SAR is of interest because it may serve as an index for comparison and extrapolation of experimental results from tissue to tissue, from animal to animal, and from animal to human exposures. It is also useful in analyzing relationships among various observed biological effects in different experimental models and subjects. This is in clear contrast to incident field or any other external measures of exposure which often can produce different induced fields and SARs inside different biological systems. Therefore, an important task in assessing the health and safety of RF energy is the determination of induced fields in biological tissues. Also, the whole-body absorption of RF electromagnetic energy by humans and laboratory animals, as specified by SA, is of interest because it is related to the energy required to alter the thermoregulatory response of the exposed subject.

Table 2.2.2.4. Quantities and units used in the radiofrequency band

Quantity	Symbol	Unit
Conductivity	σ	Siemens per meter [$\text{S}\cdot\text{m}^{-1}$]
Permittivity	ϵ	Farad per meter [$\text{F}\cdot\text{m}^{-1}$]
Current	I	Ampere [A]
Current density	J	Ampere per square meter [$\text{A}\cdot\text{m}^{-2}$]
Electric field strength	E	Volt per meter [$\text{V}\cdot\text{m}^{-1}$]
Power density	S	Watt per square meter [$\text{W}\cdot\text{m}^{-2}$]
Frequency	f	Hertz [Hz]
Impedance	Z	Ohm [Ω]
Magnetic field strength	H	Ampere per meter [$\text{A}\cdot\text{m}^{-1}$]
Propagation constant	k	per meter [m^{-1}]
Specific energy absorption	SA	Joule per kilogram [$\text{J}\cdot\text{kg}^{-1}$]
Specific energy absorption rate	SAR	Watt per kilogram [$\text{W}\cdot\text{kg}^{-1}$]
Wavelength	λ	Meter [m]

It is emphasized that the quantity of induced electric field is the primary driving force underlying the interaction of electromagnetic energy with biological systems. The induced field in biological tissue is a function of body geometry, tissue permittivity, and conductivity, and the exposure conditions. Moreover, a determination of the induced field is recommended because: (1) it relates the field to specific responses of the body, (2) it facilitates understanding of biological phenomena, and (3) it is independent of any mechanism of interaction. Once the induced field is known, quantities such as SAR can be derived from it by a simple conversion formula. For example, from an induced electric field E in V m^{-1} , the SAR can be derived as

$$SAR = \sigma E^2 / \rho_m, \quad (2.2.2.11)$$

where σ is the bulk electrical conductivity [S m^{-1}] and ρ_m is the mass density [kg m^{-3}] of tissue.

However, at present, a small isotropic, implantable electric field sensor with sufficient sensitivity is not widely available, except for cellular mobile telephone compliance testing laboratories. Consequently, a common practice in experimental dosimetry relies on the use of small field insensitive temperature probes to measure the temperature elevation produced under a short-duration (< 30 s), high-intensity exposure condition. The short duration is not enough for significant conductive or convective heat transfer to contribute to tissue temperature rise. In this case, the time rate of initial rise in temperature (slope of transient temperature response curve) can be related to SAR through a secondary procedure, i.e.,

$$SAR = c\Delta T/\Delta t , \quad (2.2.2.12)$$

where ΔT is the temperature increment [$^{\circ}\text{C}$], c is the specific heat capacity of tissue [$\text{J kg}^{-1} ^{\circ}\text{C}^{-1}$], and Δt is the duration [s] over which ΔT is measured. Thus the rise in tissue temperature during the initial transient period of RF energy absorption is linearly proportional to the value of SAR. It is important to distinguish the use of SAR and its derivation from temperature measurement. The quantity of SAR is merely a metric for energy deposition or absorption and it should not be construed to imply any mechanism of interaction, thermal or otherwise. However, it is a quantity that pertains to a macroscopic phenomenon by virtue of the use of bulk electrical conductivity in its derivation, eq. (2.2.11) and the use of specific heat capacity in eq (2.2.12).

2.2.2.5.2 Sources of concern

RFR is subdivided in different frequency bands assigned to communicate information in industry, medicine and science. Often used frequencies are 27.12 MHz (shortwave), 433 MHz (decimeter wave) and 2.45 GHz (microwave) (rms frequencies). When using other than the above described frequencies, irradiation must be limited by appropriate measures (i.e. shielding, power limitation) so that the operation of radio communication systems remains undisturbed.

2.2.2.5.2.1 Technical applications

Information technology is the main source of high-frequency and microwave radiation (radio, telecommunication, also via satellites and radar). In addition to some strong transmitter stations, smaller transmitters and mainly portable sets (mobile telephones) contribute to the steadily increasing radiation exposure of the public. Cellular mobile communication technologies have proceeded through several generations. 3G systems are the newest digital mobile communications technologies, also known as UMTS in Europe, which allows high speed data access. This and other new technologies are described in [08ICN].

From a multitude of industrial devices, those used in the welding, and drying facilities of the plastics, paper and wood industries must be considered, furthermore, most importantly facilities for heat induction in the metal processing industry. The most widely used source of microwaves besides mobile phones are microwave ovens used in households as well as in commercial facilities; other sources are microwave vulcanisation, radar air control and diathermy.

The sources may generally be divided into two groups, sources that emit their energy freely into space either as a narrow radiation beam (e.g. radar) or unbeam in all directions (e.g. radio and television stations) and, secondly, sources whereby the energy largely remains within an enclosed area (e.g. microwave ovens).

2.2.2.5.2.2 Medical applications

Diathermy

The earliest therapeutic application of radiofrequency electromagnetic fields was in diathermy. Two types of diathermy are commonly used, short-wave (usually at about 27 MHz) and microwave. Only a part of the patient's body is exposed to RF energy and exposure duration is limited (typically 15-30 minutes). However, exposure intensity is high and sufficient to cause the intended sustained increase in tissue temperature. Exposures to operators of short-wave diathermy devices may exceed $60 \text{ V}\cdot\text{m}^{-1}$ and or 0.16 A m^{-1} for operators standing in their normal positions (in front of the diathermy console) for some treatment regimes. Stronger fields are encountered close to the electrodes and cables [82Stu]. In the "worst case" high exposure of staff may occur at distances less than 1.5-2 m (27.12 MHz) or 1m (433 MHz and 2.45 GHz [84Vei].

Hyperthermia

Electromagnetic fields have also been used in inducing local hyperthermia for cancer therapy. As in diathermy, the patient is exposed to intense fields for a short time. There is relatively little information on operator exposure. One of the devices around which exposure fields have been measured operates at 13.56 MHz and employs coils for heating the torso, neck or thigh. Depending on the power to the coil and the coil type, high exposure of the operator may occur at distances of between 0.25 and 1 m from the coil.

Actually, a new brain cancer (glioblastomas) magnetic fluid hyperthermia (MFH) treatment method which has been developed by A. Jordan should be mentioned [07Mai]. Iron oxide nanoparticles of 13 nm diameter are injected directly into the tumor. A magnetic field applicator generates a 100 kHz-field. This oscillates the iron oxide particles, heating them up and thus warming the tumor to a temperature of 43° to 70° C. The tumor is completely destroyed, the side effects are minimal.

2.2.2.5.2.3 Magnetic resonance imaging

MRI is an imaging technique that employs strong DC, time-varying magnetic (gradient fields) and radiofrequency electromagnetic fields. It can image soft tissues – unobstructed by bone – with enhanced contrast. Moreover, the ability to provide images in numerous planes without requiring the repositioning of the patient has rendered MRI a very effective and important tool for soft tissue imaging. Indeed, it has become the radiological modality of choice for a great number of diagnostic procedures.

In a conventional MRI system operating at 1 T, because of its design, it is unlikely that radiological staff would be exposed to significant fields. Some newer open 0.7 T MRI systems allow medical personnel to perform interventional procedures on patients under MRI guidance. It is possible that their hands, heads or torsos may receive significant exposure under such conditions, especially for gradient fields [04ICN, 08ICN]. The gradient field is lower than the static magnetic field but it is pulsed rapidly in time and is a function of imaging technique and design of the MRI system. It is significant to note that the time rate of change of the gradient magnetic field is closely related to the strength of the electric field induced inside the body.

The demand for increased spatial resolution and high signal-to-noise ratio (SNR) from MRI instruments has prompted the use of much higher static magnetic fields (as high as 11 T). This development has led to the use of higher RF frequencies for MRI, which, in principle, not only can augment the amount of RF power deposition inside the patient's body, but also increases the EMF exposure for workers using MRI equipment in the hospital environment and workers employed for supporting, servicing, developing and manufacturing this equipment.

The issue of MRI, the protection of the patient, the personnel and general public is dealed in Sect. 5.4.1 in more detail.

2.2.2.5.2.4 RF ablation, surgery

Radiofrequency ablation is a technique that uses contact electrodes to deliver low frequency (100-500 kHz) voltages for a wide variety of medical therapies. For over a half century, an electrosurgical knife (electro surgery) has been used by surgeons to cut and cauterize tissues as a replacement for a scalpel.

Cardiac ablation uses a needle electrode, inserted through a vein, in the heart, without requiring opening of the chest wall or heart. An RF generator with a power of about 50 watts is used to creating lesions on the inner wall of the heart for the treatment of various cardiac rhythm disorders. These disorders are due to abnormal cardiac rhythms (arrhythmias) as a result of abnormal electrical pathways in the heart muscle.

Radiofrequency ablation (RFA) for cancer therapy is a new technique that uses heat to destroy cancer tumours deep within the body. A small needle electrode is placed directly into the tumour. The electrode's high frequency voltages create intense heat that can reach the boiling point of water, killing cancerous cells. This technique has been used to destroy liver tumours as well as renal and breast tumours.

RF telemetry transmitters encapsulated in a small pill have been used to monitor internal body

temperature and other physiological parameters. In addition, pills with imaging cameras have been discussed and may be developed. These devices transmit at a variety of frequencies. Devices that are planned for use in patients must pass the safety requirements of the specific countries where they are to be sold. This implies they are subject to review for excessive SAR and other concerns of the regulatory officials in each particular country where they are to be sold. Since the receiver is a few meters away (outside the body) the total radiated power from the pills does not need to be more than a few milliwatts.

2.2.2.5.3 Biophysical interaction mechanisms

The general interaction mechanisms between electromagnetic fields and the atoms, such as e.g. Zeeman splitting or Stark effect, can be studied in many physics textbooks. It is generally assumed that reactions due to spin effects are negligible because of the short life times of intermediate states in comparison with macroscopic effects such as heating. One exception is diagnostic imaging by nuclear magnetic resonance, where the effect of strong magnetic fields on the atomic structure of biomaterial is utilized. In a static magnetic field the application of high frequency magnetic fields lead to resonant energy absorption in the sample which depends on field strengths, temperature and frequency.

Electric and magnetic fields may exert forces on the charged particles in materials, thus altering the original charge patterns. These altered charge patterns in the materials produce local electric and magnetic fields in addition to the external fields. In non-magnetic materials, mainly the applied electric field has an effect on the charges in the material. This occurs in three primary ways: polarisation (e.g. induction of dipole moments), orientation of permanent dipoles, and drift and diffusion of conduction charges (both electronic and ionic). Materials primarily affected by the first two effects are called dielectrics; materials primarily affected by the third one are termed conductors. Most biological material is dielectric, and shows the above effects. Biomaterial also is a conductor, but a very weak one.

Many effective mechanisms can be comprehended on a molecular and cellular level by examining the macroscopic electric properties of the material (dielectric constant and electric conductivity) [63Sch, 87Pet, 86Fos, 09ICN1].

Orientation polarization is the main cause of energy absorption in biological aqueous substances for frequencies between 1 and approximately 100 GHz. The maximum energy dissipation of free water molecules is at about 20 GHz. Bound water, peptides, proteins or side chains of larger molecules have their maximum absorption at frequencies between 1 and 20 GHz. The orientation movements do not lead to molecular changes; the absorbed energy is completely transformed into heat.

Another mechanism occurs mainly in the frequency range between 100 kHz and 100 MHz and can be described as a bounding layer polarization frequently occurring in structures with different electric properties. Due to the poor electric conductivity of the cell membrane, in comparison to the interior and exterior cell space, the electric high-frequency current produced by the electromagnetic field in tissue essentially flows around the cell for lower frequencies (e.g. 10 kHz). Due to charge displacements, the cell acts here like a large dipole. At higher frequencies, on account of the capacity of the cell membrane, a large part of the high-frequency current then flows through cell membrane and cytoplasm.

Charge displacements around and within a biological cell result in a difference of the electric potential between the internal and external space of the cell, which superimposes the existing resting potential of the membrane and shows the same time dependence as the external field. Since cells behave similarly to dipoles, these field-induced dipoles transfer forces to other dipoles in their environment, as was verified by appropriate experiments [77Sch].

The interaction mechanisms on an atomic and molecular level and the subsequent effects on the microscopic level – heat effects, the generation of electrical potential difference and field-induced force effects – are not independent from each other and are capable of occurring simultaneously according to frequency and exterior field strength, as shown by *in vitro* studies. Under given physiological conditions the heat effect is predominant in most cases. For molecular structures, further quantum-mechanical or cooperative absorption effects are postulated (see reviews in [86Pol]) but bear no significance for risk assessments.

On a macroscopic level, additional factors must be considered. Permeability and dispersion, absorption and transformation of radiofrequency energy in a biological object depend not only on the

various layers with different dielectric properties but also on form and position of the object in relation to the external electromagnetic field. A considerable degree of simplification is required for realistic radiation purposes.

The penetration depth of high-frequency radiation can be derived from the frequency-dependent dielectric tissue data. For muscle tissue, the penetration depth (distance over which about $\frac{2}{3}$ of the radiation energy is absorbed) measures about 10 cm at 30 MHz and 3 cm at 1 GHz [80Sch]. Microwaves practically behave like infrared radiation above 20 GHz.

The absorbed energy can be calculated from [eq. \(2.2.2.11\)](#). Variously structured and combined models are available from calculations and experimental studies, beginning with dosimetry handbooks by Durney et al. 1978 (e.g., [86Dur]), until more recent modern numerical techniques (e.g., [97Dim, 02Dim, 05Dim]). The dosimetry handbook shows the essential principles of the dependence of energy absorption on exposure conditions and frequency. Almost all absorption curves apply to the far field, and thus are inappropriate for dosimetric purposes in the near-field range. For exposure situations in the near-field modern fine resolution FDTD calculations in anatomical realistic voxel models of the human body can be applied.

The frequency-dependent SAR curves can be divided into three sections. In the lower frequency range (sub resonance range below 30 MHz) the absorption of energy decreases rapidly at decreasing frequency. In order to produce heat in the body, the power flux densities must be high. The value for the exposure limit may be higher in this frequency range than in others. The distribution of absorbed energy or induced body field strength is, on the other hand, particularly inhomogeneous. Non-thermal membrane effects are, in addition, predominant at frequencies below 100 kHz [79ber].

In the resonance range (about 30-400 MHz), object (whole body or body parts) and wavelength are of similar order. High absorption is possible, and exposure limits must be set at lower values. The resonance frequency range is higher for children than for adults, due to the smaller body configuration of the former. Shifts in resonance frequency are not only due to body dimensions, but also to the influence of conducting and body-connected surfaces. These factors move the resonance frequency to lower values.

In the upper frequency range, the wavelength is shorter in comparison to the object. The penetration depth is low. An increased frequency will only heat up surface layers. In the frequency range between 200 and 3000 MHz refraction may furthermore result in focusing effects producing spatially limited "hot spots" in the body. For example, a considerable rise in temperature may occur in certain areas of the head [75Kri]. This effect may explain the phenomenon of being able to audibly perceive modulated high-frequency radiation [74Fosl].

The various absorption curves for adults, children and infants can be summarized by an enveloping SAR curve, which serves as a basis for most exposure limits calculated according to the SAR concept (e.g. [98ICN]).

The SAR curves for animals (e.g. mice, rats) differ by one to two orders of magnitude from human SAR curves where resonance frequency and SAR values at identical frequencies are concerned. This must be taken into consideration when extrapolating results from animal experiments to man.

For determining a temperature increase it is important to observe the cooling of tissue from heat conduction, heat radiation, transport via blood and perspiration. A power density of $100 \text{ W} \cdot \text{m}^{-2}$ increases the average temperature from 0.5 to 2° C (depending on frequency), taking heat transport and normal blood flow into consideration. Unfavourable ambient temperature and humidity lower the maximum tolerable thermal stress. Conditions that impair thermoregulatory functions include, for example, fever, diabetes, cardiovascular disease, obesity, old age and special medications. At a temperature increase of about 0.5 °C, the normal thermoregulation in the human hypothalamus begins to respond.

In poorly circulated organs, such as the eye lens, the temperature increase is greater; power densities of more than $1 \text{ kW} \cdot \text{m}^{-2}$ may lead to the formation of a cataract [79Car]. Accordingly, the eyes must be protected during HF therapy in the head region or the applied HF power must be reduced.

For hazard assessment indirect field coupling may be more important than direct field coupling. People with metallic implants or active medical devices such as pacemakers or insulin pumps, as well as patients in intensive care units of hospitals may be at risk due to the malfunctioning of these medical devices if exposed to electromagnetic fields. Malfunctioning of pacemakers due to interference at the work place was described for electric arc welding, transmitting towers, dielectric heaters and diathermy or hyperthermia devices. A completely different energy distribution may result if metallic implants are

present in the body. For large implants, locally increased high-frequency currents in the tissue may produce burns on the contact surfaces of metal-tissue. Effects such as this must be taken into consideration for RF therapy.

2.2.2.5.4 Summary

Radiofrequency radiation is subdivided in different frequency bands assigned to communicate information in industry, medicine and science. Information technology is the main source of high-frequency and microwave radiation (radio, telecommunication, also via satellites and radar). In addition to some strong transmitter stations, smaller transmitters and mainly portable sets (mobile telephones) contribute to the steadily increasing radiation exposure of the public. From a multitude of industrial devices, those used in the welding, and drying facilities of the plastics, paper and wood industries must be considered, furthermore, most importantly facilities for heat induction in the metal processing industry. The most widely used source of microwaves besides mobile phones are microwave ovens used in households as well as in commercial facilities; other sources are microwave vulcanisation, radar air control and diathermy. Medical applications include diathermy and hyperthermia, nuclear magnetic resonance (MRI), surgery and RF ablation.

The biophysical interaction mechanisms of electromagnetic fields are not independent from each other and are capable of occurring simultaneously according to frequency and exterior field strength. The following general statements can be made concerning risk assessment, taking into account thermal and athermal effects:

- <100 kHz: only the values of internal electric fields or internal current densities are important.
- 100 kHz-10 MHz: current densities and specific absorbed radiation (SAR) have to be taken into account (overlap region).
- < 10 MHz: electric and magnetic fields may be considered separately.
- > 10 MHz: electromagnetic waves are present, only the specific absorbed radiation (SAR) has to be considered.
- < 100 MHz: contact currents may be significant up to that level.
- 100 MHz: break down of membrane potential.
- Force effects on cells are not significant; they are mostly dominated by other effects.
- Interference with medical devices is possible in the whole frequency range.

The biological effects are explained in detail in [Sect. 2.3.2.5.2](#).

2.2.2.6 Optical radiation and lasers

2.2.2.6.1 Introduction

Electromagnetic radiation with wavelengths in the range between 100 nm and 1 mm is widely termed optical radiation. Ultraviolet radiation (UV) is in the wavelength range between 100 nm and 400 nm, light (or visible radiation) from 400 to 760 nm and infrared (IR) radiation from 760 nm to 1 mm. The UV and IR regions may be further subdivided as shown in [Table 2.2.2.1](#). While these spectral bands are useful "short-hand notations" for discussing the photobiological effects of optical radiation, the predominant biological effects have less sharply defined spectral limits. Because there is little energy emitted by conventional light sources in the far-infrared (IR-C range), only lasers in this spectral region pose potential hazards.

The special optical properties of lasers vary significantly from those of conventional broad-band optical sources, and so they are treated separately. Unlike lasers, which generally emit at one wavelength, spectroradiometric data, measurements and geometrical factors must be considered when evaluating potential hazards of conventional light sources.

Although this book will not deal with medical applications of optical radiation like diagnostic purposes in dentistry, UV phototherapy (PUVA), photo chemotherapy, photodynamic therapy or IR hyperthermia, it

seems convenient to give the relevant information of optical radiation to the medical physicist. For further reading see, for example, the following reviews: [80Sli, 82WHO, 02AGN].

2.2.2.6.2 Radiometric terms and units

Prior to any meaningful determination of the optical radiation exposure of biological tissues, it is necessary to define the relevant quantities and units. For all photobiological effects, it is necessary to measure the appropriate radiometric quantity. The surface exposure dose rate is termed the *irradiance*, with units of watts-per-square-centimeter [W cm^{-2}], and the surface exposure dose is termed the *radiant exposure*, with units of joules-per-square-centimeter [J cm^{-2}].

There are also parallel dose rate and dose concepts within scattering tissue, and these quantities are termed *fluence rate*, also with units of watts-per-square-centimeter [W cm^{-2}], and dose within tissue that is termed the *fluence*, also with units of joules-per-square-centimeter [J cm^{-2}]. The existence of two terms for the same radiometric unit seems curious, and this has confused many scientists, with the result that the terms are frequently misused for the other. But the concepts are different and the distinctions are important. The quantities irradiance and radiant exposure are what instruments measure at the exposed surface (and follow Lamberts Cosine Law), but fluence rate and fluence include backscattered light and are useful for photochemical calculations within tissue (as in photodynamic therapy).

Table 2.2.2.5. Useful Radiometric Units

Term	Symbol	Definition	Unit
Radiant Energy	Q	Energy emitted, transferred, or received in the form of radiation	joule [J]
Radiant Power	P	Radiant energy per unit time	watt [W] defined as [J s^{-1}]
Radiant Exposure (Dose in Photobiology)	H	Energy per unit area incident upon a given surface	joules per square centimeter [J cm^{-2}]
Irradiance or Radiant Flux Density (Dose Rate in Photobiology)	E	Power per unit area incident upon a given surface	watts per square centimeter [W cm^{-2}]
Integrated Radiant Intensity	I_P	Radiant energy emitted by a source per unit solid angle	joules per steradian [J sr^{-1}]
Radiant Intensity	I	Radiant power emitted by a source per unit solid angle	watts per steradian [W sr^{-1}]
Integrated Radiance	L_P	Radiant energy emitted by a source per unit solid angle per source area	joules per steradian per square centimeter [$\text{J sr}^{-1} \text{cm}^{-2}$]
Radiance	L	Radiant power emitted by a source per unit solid angle per source area	watts per steradian per square centimeter [$\text{W sr}^{-1} \text{cm}^{-2}$]

Photobiological quantities

In photobiology, the concept of a biologically effective dose is of critical importance. Since not all wavelengths of optical radiation are equally effective in producing a biological effect, an action spectrum $A(\lambda)$, which defines the relative effectiveness of different wavelengths, is determined. This relative response curve is generally normalized to provide a maximal value of 1.0 at the wavelength of maximal tissue sensitivity. When considering health effects of optical radiation, an effective exposure rate (i.e., irradiance) E_{eff} (or the exposure summed over time, i.e., the effective radiant exposure H_{eff}) is calculated

by spectral weighting as follows: the spectral irradiance E_λ at the surface of the exposed biological tissue is mathematically weighted against the action spectrum of the biological response $A(\lambda)$ across the relevant spectrum and is shown as follows:

$$E_{\text{eff}} = \sum E_\lambda A(\lambda) \Delta\lambda \quad (2.2.2.13)$$

The effective exposure dose (or effective radiant exposure) H_{eff} is the product of the exposure duration t , in seconds, and the effective irradiance E_{eff} (spectrally weighted optical radiation):

$$E_{\text{eff}} (\text{in W cm}^{-2}) t = H_{\text{eff}} (\text{in J cm}^{-2}) \quad (2.2.2.14)$$

[Eq. \(2.2.2.14\)](#) can also be rearranged to calculate the exposure time t necessary to reach a reference exposure dose for the given response:

$$t = H_{\text{eff}} (\text{in J cm}^{-2}) / E_{\text{eff}} (\text{in W cm}^{-2}) \quad (2.2.2.15)$$

For example, the exposure duration t_{erythema} necessary to achieve a minimum erythemal dose (MED) in an individual would be the MED for that individual, e.g., 220 J m^{-2} divided by the erythemally weighted irradiance E_{erythema} , (weighted against the appropriate erythemal action spectrum $E(\lambda)$).

As another example, the exposure guidelines that are referred to in the section below apply an analogous formula using a different spectral weighting, a hazard action spectrum, $S(\lambda)$.

2.2.2.6.3 Optical radiation sources

Solar radiation

Sunlight has played a critical role in the development of life on Earth. The infrared and visible regions of the solar radiation comprise 95% of the total radiation reaching the Earth's surface. The ultraviolet component of the terrestrial solar spectrum comprises 5% of the radiant energy; however this component is largely responsible for the deleterious effects of solar radiation.

For further details of the characteristics of solar radiation the reader is referred to the special literature.

Artificial sources: ultraviolet radiation (UVR)

Sources of optical radiation can be characterized by an arc discharge (e.g. welding arc, metal halide lamp), incandescent lamps (e.g. tungsten halogen lamp) and lasers (e.g. excimer laser). Artificial sources may provide additional exposure that may be elective (e.g. sunbathing, cosmetic tanning with sunbeds, or medical therapy) or as a consequence of occupation (e.g. electric arc welders) [[88McK](#)].

Low pressure mercury-discharge (*Germicidal*) lamps are often used for the purpose of disinfection. Such lamps are very efficient emitters of 254 nm radiation (UVC). This type of lamp emits minimal amounts of visible and infrared radiation.

The most common application of the low-pressure discharge is the *fluorescent lamp*. Light is produced by conversion of the 254 nm mercury emission to longer wavelength radiation by means of a phosphor coating on the inside of the glass envelope of the lamp. Lamps are available with many different phosphors and envelopes to produce a wide range of spectral emissions covering the visible (light), UVA and UVB regions. While the continuum emissions of fluorescent lamps are characteristic of the phosphors, the narrow peak, spectral emissions are dominated by the characteristic line emission spectrum of the low-pressure mercury vapour discharge.

General lighting fluorescent lamps, intended for general lighting purposes, are available in a range of physical sizes, powers and phosphors to emit visible radiant energy. The range of phosphors includes a large selection of "near white" and "special colour" lamps. A plastic cover (diffuser) placed over the lamp eliminates effectively UVR.

Metal halide and mercury lamps along with medium pressure mercury vapour lamps are used for general illumination and for health-care applications, including the phototherapy of skin diseases. The spectral emissions of the discharge are in the visible spectrum (blue, green and yellow) and a large amount of UVR is also generated. The addition of metal halide to the mercury vapour enhances the emission in the UVR.

The spectral emission of *xenon lamps*, closely matches that of sunlight for wavelengths shorter than the infrared (760 nm). This enables their use as solar radiation simulators, for example, in investigating patients with skin diseases induced or aggravated by sunlight. Large amounts of UVA, UVB and UVC are emitted by unfiltered lamps, to the extent that they can present a significant health hazard if incorrectly used. Xenon lamps are also used in high intensity endoscopic illuminators.

Quartz halogen (or *tungsten halogen*) lamps are widely used in special illumination applications, e.g., for specialized task lighting demanding high localized illumination and in instruments such as endoscopes. The quartz envelope permits the emission of UVR that may present a hazard in some circumstances.

Electrical *welding arcs* produce hazardous levels of UVR that depend upon the arc current, the shielding gas and the metal being welded.

Ultraviolet lasers and light emitting diodes (LEDs): Lasers operating in the UVR spectral region are used in medical environments for diagnostic and treatment procedures. For example, the argon fluoride laser operating at 193 nm is commonly used for corneal refractive surgery procedures. UVR emitting LEDs are relatively new UVR sources (solid-state miniature lamps) with growing applications. LEDs operating in the UVR are used in industry and research for photobiological, fluorescence detection, and materials research.

Artificial sources: visible and infrared radiation

The optical radiation from an *incandescent lamp* results from the heating of a tungsten filament to temperatures approximately 2500 to 3000 K with a maximum spectral output in the near infrared (IRA) region. Incandescent lamps are the oldest types of lamp which are still used for general lighting mainly in home environment. Tungsten halogen lamps are special types of incandescent lamps used when spotlighting or very high powers are needed.

Infrared radiation has been widely used in physical medicine for treatment of sports injuries, muscle aches, pain, and some chronic diseases. In recent years there has been an interest in the use of IR-A sources for hyperthermic treatment of cancers [01Weh].

Because of the deeper penetration of IR-A, this is used almost exclusively, and water filtering of IR to achieve pure IR-A has been recommended in therapeutics. The typical treatment irradiance of several therapeutic IR devices falls in the range of 800 W m^{-2} . Monitoring rectal temperature is the most basic indicator of core body temperature.

Along with conventional lighting, *fluorescent lamps* with spectra simulating the ambient daylight are used for mental photostimulation, e.g. to eliminate tiredness during night work, or for light therapy at specialised clinics. Light therapy has become popular for decreasing symptoms suspected to be caused by darkness, especially during the northern winter season.

Special fluorescent lamps are also used for medical treatment of hyperbilirubinaemia (neonatal jaundice) in neonatal intensive care units. Visible wavelengths less than 470 nm are the most effective in reducing serum bilirubin levels.

Metal halide lamps are discharge lamps which, in addition to mercury vapour, contain different metals in their halide salts. The colour temperature is typically 5600 K, and the resulting spectrum contains emission lines over a wide range of wavelengths, including strong peaks in the blue-light region. They are used for general lighting in factories and shops, and for floodlighting at television work.

Artificial sources: laser radiation

Lasers have contributed enormously to the advancement on evolution of medicine. The two most common and conventional high-powered operating-room lasers are the Nd:YAG and CO₂. The argon and frequency-doubled YAG-KTP lasers both produce a green wavelength of light that can be delivered through an optical fibre. The dye laser has been used to remove stones lodged in the urinary tract and so on. Low power diode laser and He-Ne laser have been used in biostimulation. In the past decade, excimer lasers have gained publicity because of their potential usefulness in medical application. The great advantage of these short-pulse, high-energy lasers is their ability to ablate tissue with minimum thermal damage. The most of current procedure employ XeCl pulsed laser ablation with the over-the-guidewire technique as a guidance. Recently some researchers use the Ho:YAG and Er:YAG laser for clinics. Short pulses from the Er:YAG have an ablative effect with little thermal damage; and also the Er:YAG offers a

superior cutting tool when longer pulses are used. The pulse laser ablation can offer restricted thermal damaged layer thickness which is independent from irradiation period.

The another remarkable advances and emerging technologies have been appeared in research and developments of high-power diode laser. They have already achieved the maximum output to approximately 60 W. Some manufactures developed the medical systems based on the tuneable laser crystal, titanium-doped sapphire. Frequency-conversion techniques, including a proprietary method of extending the range in the infrared, are expected to yield a device that tunes from 325-3000 nm. This widely tuneable laser may be the basis for a universal surgical tool that suits a variety of applications. The free-electron laser (FEL) may also find use in the future world of medicine.

2.2.2.6.4 Mechanisms of interaction

Tissue interactions in the optical spectral range are either thermally or photochemically initiated. The optical hazards of intense light sources, such as welding arcs, some tungsten-halogen lamps, and lasers can be grouped into several types of hazards to the eye and skin [80Sli, 82WHO, 94WHO, 97ICN, 99CIE, 00ICN] and are further explained in Sect. 3.3.2.5.2.

Skin cancer caused by optical radiation sources in the absence of ultraviolet radiation is not considered to be a significant risk [92IAR, 94WHO].

Characteristics of photochemical interaction mechanisms

The threshold dose for photochemical injury is the product of the dose-rate and the exposure duration. It is subject to the principle of reciprocity (the Bunsen-Roscoe Law of Photobiology); thus, for example blue-light retinal injury (photoretinitis) can result from viewing either an extremely bright light for a short time or a less bright light for longer. Reciprocity helps to distinguish these effects from thermal injuries (see below). For photochemical injury of the retina, the action spectrum peaks at approximately 440 nm for the eye with an intact crystalline lens (i.e. the phakic eye).

Characteristics of thermal interaction mechanisms

Unlike photochemical injury, thermal injury does not show reciprocity between irradiance and exposure duration. Thermal injury is strongly dependent upon heat conduction from the irradiated tissue. It requires an intense exposure within seconds to cause tissue coagulation; when exposure is less intense, surrounding tissue conducts heat away from the exposed site. Thresholds for acute thermal injury of both cornea and retina in experimental animals have been corroborated for the human eye by flash burn accident data. Normally a temperature of at least 45°C is necessary to produce a thermal burn; higher temperatures are required for thermal injury to result from exposures of shorter duration [89All1]. The irradiance required to achieve these temperatures depends upon the ambient tissue temperature and the exposure spot size. Because of the more efficient cooling of small spots, injury of small spots requires higher irradiances than injury of large spots. This more rapid cooling of small images also limits the duration of elevated temperature after the cessation of optical exposure, therefore influencing the critically important time-temperature history of the exposed tissue. Thus there is no single critical temperature for a given exposure duration; the spot size must be specified for defining ocular or skin exposures.

Laser radiation

Laser biological effects are the result of one or more competing biophysical interaction mechanisms – thermal, acoustic, optical (electrolytic breakdown), and photochemical – which vary depending upon spectral region and exposure duration. For example, in the 400-1400 nm bands, thermal injury to the retina resulting from temperature elevation in the pigmented epithelium is the principal effect for exposure durations less than 10 s and superficial thermal injury to the cornea and skin occurs at wavelengths greater than 1400 nm. Thermoacoustic injury occurs at pulse durations less than approximately 0.1 ms and can lead, for example, to haemorrhagic lesions of the retina from Q-switched lasers. Optical breakdown and plasma formation become important only from sub-nanosecond exposures. Photochemical injury

predominates in the UV spectral region and is also the principal type of injury resulting from lengthy exposures (10 s or more) to short-wavelength visible radiation (principally "blue light").

The biological effects in the different optical spectral ranges, including lasers, are explained in detail in Sects. 2.3.2.6.2–2.3.2.6.4.

2.2.2.6.5 Summary

Electromagnetic radiation with wavelengths in the range between 100 nm and 1 mm is widely termed optical radiation. Ultraviolet radiation (UV) is in the wavelength range between 100 nm and 400 nm, light (or visible radiation) from 400 to 760 nm and infrared (IR) radiation from 760 nm to 1 mm. Besides the natural solar radiation there are numerous artificial sources in industry, research and medicine. Sources of optical radiation can be characterized by an arc discharge (e.g. welding arc, metal halide lamp), incandescent lamps (e.g. tungsten halogen lamp) and lasers. Tissue interactions in the optical spectral range are either thermally or photochemically initiated. The optical hazards of intense light sources, such as welding arcs, some tungsten-halogen lamps, and lasers can be grouped into several separate types of hazards to the eye and skin, which are dependent on the wavelength. For risk assessment, the biologically effective dose is of critical importance. When considering health effects of optical radiation, an effective exposure rate (i.e., irradiance) at the surface of the exposed biological tissue is mathematically weighted against the action spectrum of the biological response across the relevant spectrum.

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2.3.1 Biological Effects of Ionizing Radiations

Radiation and Biological Effects Biological Effects

A. KAUL

In Volume VIII/4 “Radiological Protection” of Landolt-Börnstein, published in 2005 [05Kau], H. Smith and J.W. Stather have described in Chapter 2 in great detail the biological effects of ionizing radiation on the body and body organs. The Chapter covers both effects of the subcellular and cellular level from both acute exposure from high radiation doses and/or dose rates which can cause severe tissue damage, or from irradiation of man to low doses and/or dose rates which can give rise to radiation induced cancer and hereditary disease in future generations. Exposures of man from high radiation doses and/or dose rates are usually those in medical therapeutic radiology; those of low doses and dose rates are generally in diagnostic radiology, usually with the exception of interventional radiodiagnostics.

Since Volume VIII/4 “Radiological Protection” of Landolt-Börnstein has been edited quite recently, and addresses to both the qualified experts in radiological protection, medical radiation physics and to physicians practising in X-ray diagnostics, nuclear medicine and radiation oncology, the reader is referred to Chapter 2 of Landolt-Börnstein Volume VIII/4 [05Kau]. Most recent information on biological effects of ionizing radiations has been published by the ICRP in its Publication 103 [07ICR] as a basis for revision of its Recommendations of 2007. Consequently, based on Vol. VIII/4 and ICRP Publ. 103 the following Sections summarise as a basis for better understanding the biological background of the subsequent Chapters of the present Volume the most important information on biological effects of ionizing radiations, and of possible risks for the different biological effects of these radiations. This information specifically aims at those readers whose major professional and scientific interest is devoted to medical radiological protection and to medical physics in radiooncology, the latter as the major subject of Volume VIII/7B, to be published later.

As mentioned above, the Sections below are mainly based on the text of H. Smith and J.W. Stather as well as on ICRP Publ. 103. They partially adopt formulations and sometimes take over whole paragraphs of the above authors and ICRP Publ. 103. For more detailed information the reader is referred to the original Chapter of the authors in Landolt-Börnstein Volume VIII/4 [05Kau], and to the relevant Chapters of Publication 103 of the ICRP [07ICR].

2.3.1.1 Introduction

The timescale of events leading to radiation effects following exposure to ionizing radiations can be summarised as follows. Ionizations (and excitations; timescale 10^{-20} - 10^{-14} s) are followed by physico-chemical events (formation of radicals and radical interactions). The repair of damage may be completed within tens of minutes while effects in cells can arise within hours or days. The biological manifestations in multi-cellular organisms, including man, can be delayed for many years or, as in the case of hereditary disease, only be manifest in future generations.

The widespread use of X-rays and radium in treating disease in the early 1900s led to the recognition of a cancer risk in organs and tissues following high radiation doses. However, there was a delay of about 40 years before it became clear that there was a risk of radiation-induced cancer from irradiation at lower doses, too, without any apparent threshold dose. In addition animal studies have shown an increased incidence of certain types of inherited disorders in the descendants of irradiated parents. For both these biological effects – cancer and inherited disorders – the probability of their occurrence, but not their severity, depends on the radiation dose. In radiological protection terminology these effects are termed *stochastic effects*.

A second type of damage may be seen quite soon after exposure of the whole body or of parts to high radiation doses in the order of few Gy or tens of Gy, the severity of which is related to the extent of radiation exposure. In addition from observations it is assumed that there is a threshold below which the

clinically detectable damage does not occur. Due to the obvious impairment of the functional capacity of tissues these effects are referred to as *deterministic effects*.

2.3.1.2 Deterministic effects

2.3.1.2.1 Dose response relationships for radiation damage

The probability of detecting loss of tissue or organ function following exposure to ionizing radiation increases steeply above a threshold dose to a maximum. Above the threshold dose, the severity of the effect also increases with dose reflecting more cell loss and hence damage to tissue function. There is individual variation in radiosensitivity in any exposed population reflecting differences in the ability of individuals to cope with radiation-induced cellular damage.

The reason for the presence of this threshold dose is that radiation damage (serious malfunction or death) of a critical population of cells in a given tissue needs to be sustained before injury is expressed in a clinically relevant form.

2.3.1.2.2 Whole body irradiation

Evidence of deterministic effects of ionizing radiation comes from retrospective studies of radiotherapy patients, radiologists in the early years of therapeutic administration of X-rays and natural radioactive substances – especially radium – after Wilhelm Conrad Roentgen's discovery of X-rays, and of natural radioactivity by Henri Becquerel. More further information from Japanese populations exposed to ionizing radiations from the explosions of the atomic bombs in Hiroshima and Nagasaki, and from individuals accidentally exposed to high doses following nuclear reactor accidents and radiographic sources.

Early (days to weeks) tissue reactions to radiation in cases where threshold doses have been exceeded may be of the inflammatory type resulting from the release of cellular factors, or they may be reactions resulting from cell loss. Late tissue reactions (months to years) can be of generic type if they arise as a direct result of damage of that tissue.

After exposure to doses of a few Gy, the depression in the numbers of circulating white blood cells and blood platelets may be severe as to result in death from infection and haemorrhage. Depression of the stem cells providing the protective mucosal cells lining the intestinal tract wall results in a denuding of the gut surface (gastrointestinal syndrome). Further severe acute radiation effects from whole body exposures in the order of several Gy are leakage of blood from damaged blood capillaries, the haematopoietic syndrome, damage to endothelial cells lining the alveolar air sacs (which may occur after radiotherapy and after the inhalation of high specific activity radioactive particles). After higher whole body doses in the order of above 15 Gy coma and death develops rapidly after exposure.

2.3.1.2.3 Partial body irradiation: tolerance dose

Intensive experience in the treatment of patients undergoing radiotherapy has provided data upon which to determine the tolerability of healthy tissues and organs to ionizing radiations. The term tolerance dose in adults after radiotherapy is defined as the amount of radiation received during conventional treatment below which unacceptable effects do not occur in more than a few percent of patients within 5 years following treatment. The tolerance doses for some adult tissues are in the order of 1-2 Gy for total bone marrow (blood cell depletion), 2-6 Gy (ovaries; permanent sterilisation), 3-4 Gy (testes; permanent sterility), 5-10 Gy (eyes; cataract), 40 Gy (lung; pneumonitis), >45 Gy (thyroid; hypoplasia).

In contrast to the precise exposure conditions of radiotherapy, exposure of workers to high doses of low-LET radiations is most likely to be non-uniform and resulting from mixed radiations. The tolerance dose therefore can at best be used as a cautious approximation to a threshold dose. Recommended values

by ICRP for the most radiosensitive tissue and organs are given in [91ICR]. For equivalent dose brief (Sv) and protracted exposure (Sv yr^{-1}) the corresponding values are in the order of 0.15 Sv (testes, temporary sterility) to 6 Sv (ovaries, sterility) for the brief exposure, and 0.01 Sv yr^{-1} (lens, detectable opacities) to 2 Sv yr^{-1} (testes, permanent sterility) for protracted exposure. The threshold dose for death is in the range of 6 to 7 Gy if the radiation is spread over 30 fractions in a period of 6 weeks.

Based upon extensive experience in the use of fractionated x- and gamma radiation in radiotherapy (typically 20-30 fractions each of 2 to 6 Gy over several weeks) various degrees of skin damage can be observed according to the area, depth of skin involved, absorbed dose, duration and frequency of exposure. The earliest observable change of a transient reddening of the skin within a few hours after exposure to dose is above about 2 Gy.

2.3.1.2.4 Irradiation in utero

For the developing embryo and fetus there is evidence that deterministic effects such as malformations of the brain, skeleton, eyes and heart as well as severe mental retardations may occur following irradiation in utero. Evidence of those deterministic effects of ionizing radiations on the embryo and fetus is derived almost entirely from animal experiments. From extrapolations of these results the following human threshold doses for radiological protection purposes for low-LET radiation were proposed by ICRP for use with utmost caution [07ICR]: In respect of the induction of malformations, the new data strengthen the view that there are gestational age-dependent patterns of the in-utero radiosensitivity with maximum sensitivity being expressed during the period of major organogenesis. On the basis of animal data it is judged that there is a true dose threshold of around 100 mGy for the induction of malformations.

A study of about 1600 children exposed in utero at Hiroshima and Nagasaki to various radiation doses and at various developmental stages has shown about 30 cases of clinically severe mental retardation with a greater incidence than expected in higher dose groups. Maximum mental retardation was observed exposed between 8 and 15 weeks after conception and somewhat lower between 16 and 25 weeks with no effect following exposures later than 25 weeks [84Ota; 88Ota]. Although the number of cases is small, the data indicate an excess probability of mental retardation of 40% at 1 Sv received during the 8-15 weeks after conception. At doses in the order of 0.05 Sv, no effect would be detectable in the general distribution of the IQ as a classification index of mental retardation. The associated data on IQ losses were estimated at around 25 points per Gy, however, they are difficult to interpret, although a non-threshold dose response cannot be excluded.

2.3.1.3 Stochastic effects

2.3.1.3.1 Cancer induction and development

The development of cancer in tissues can be subdivided into the phases neoplastic initiation, neoplastic promotion, conversion and progression. Neoplastic initiation encompasses irreversible cellular damage which provides the potential in cells for the development of cancer. This initiation process results from damage to DNA leading to gene mutations in single target cells in tissues. Promotion is the process of a subsequent abnormal growth of cells in semi-independent manner to become fully malignant in a process of neoplastic development. Subsequent progression is characterised by invasion of these malignant cells into adjacent normal tissues with final establishment of metastases at other sites in the body.

Thus, a single mutational event in a critical gene in a single target cell *in vivo* can create the potential for neoplastic development. Consequently a single radiation track traversing the nucleus of a target cell has a finite albeit very low probability of generating the specific damage to DNA, which means at the level of DNA damage that there is no basis for assuming a dose threshold below which the risk of tumour induction by ionizing radiations would be zero. For radiation protection purposes a progressive increase in risk with increasing dose with no threshold is therefore assumed [95Cox]. Whilst a multistage mechanism as described above is considered to be the cause of many human tumours, there are likely

some tumours that may arise in tissues where there has been deterministic damage for such tumour types, so that a threshold dose may need to be expected before the tumour will appear. For example the development of radiation-induced bone tumours in former radium dial painters (internal exposure to alpha-emitters) may require a threshold dose due to deterministic damage to be exceeded [00UNS].

2.3.1.3.2 Risk of cancer

These advances in knowledge of the cancer process in general give increased confidence that detailed information on DNA damage - response/repair and induction of gene/chromosomal mutations - can contribute significantly to judgements on the radiation-associated increase in the incidence of cancer at low doses. This knowledge also influences judgements on relative biological effectiveness (RBE), radiation weighting factors w_R , tissue weighting factors w_T , and dose and dose rate effects (see Sect. 2.3.1.4).

2.3.1.3.3 Dose-response relationships, risk estimation and the radiological concept of limiting the risk

The total radiation dose and the dose rate both influence cancer induction and are linked to the form of the dose-response relationship. As indicated above for radiological protection purposes tumour induction is generally assumed to increase with increasing dose with no threshold. At very low doses where there is a low probability of more than one radiation event occurring in a cell nucleus it may be expected that the effect is linearly related to dose. At higher dose where multiple ionizing events within a single cell are commonplace, damage arising from interactions between two or more events becomes more probable, so that at high doses cell killing will progressively reduce the risk of tumour induction. An approach commonly used in risk assessment is to fit a linear dose-response relationship to the data of cancer incidence at higher doses, a procedure usually considered to give an upper limit to the risk at low doses.

The ICRP judges that the weight of evidence on fundamental cellular processes coupled with dose-response data supports the view that, in the low-dose range, below about 100 mSv, it is significantly plausible to assume that the incidence of cancer or heritable effects (see Sect. 2.3.1.3.4) will rise in direct proportion to an increase in the equivalent dose in the relevant organs and tissues. This dose-response model is generally known as "linear-non-threshold" LNT. Its adoption combined with a judged value of a dose and dose rate effectiveness factor provides a prudent basis for the practical purposes of radiological protection, i.e. the management of risks from low-dose radiation exposure.

The data on the A-bomb survivors provide information on risks from radiation-induced cancer in a range of tissues, although today no quantitative information is available from these human populations for radiation-induced cancers of the liver, cells on the bone surfaces, thyroid and skin. Information on radiation-induced cancer in these tissues is, however, available from other epidemiological studies in populations of medical diagnosis (e.g. multiple fluoroscopies or thorotrast patients), medical therapy (e.g. pelvic radiotherapy or neck and chest radiotherapy), occupational exposure (e.g. uranium miners or radium dial painters), and radiation accidents (e.g. radiation workers or populations exposed by Chernobyl reactor accident fall-out). From these investigations international governmental and non-governmental organisations such as UNSCEAR, BEIR, ICRP calculated risks of radiation-induced cancer for different populations (lifetime fatal cancer risks) and risk coefficients for fatal organ or tissue cancer. These risk coefficients have been used by ICRP in developing the dose limits given in their recommendations and provide the basis for the International and European Basic Safety Standards (for more information see ICRP Publication 103 [07ICR] and Chapter 5 of the present Volume).

2.3.1.3.4 Risk of heritable effects

There continues to be no direct evidence that exposure of parents to radiation leads to excess of heritable disease in offspring. The ICRPs [07ICR] present estimate of genetic risks up to the second generation of about 0.2% per Gy is essentially the same as that cited by UNCEAR [01UNS] in 2001.

2.3.1.3.5 Induction of diseases other than cancer

Evidence has accumulated that the frequency of non-cancer diseases is increased in some irradiated populations. The strongest statistical evidence for the induction of these non-cancer effects at effective doses in the order of 1 Sv derives from the most recent mortality analysis of the Japanese atomic bomb survivors followed after 1968 [03Pre]. That study has strengthened the statistical evidence for an association with dose particularly for heart disease, stroke, digestive disorders, and respiratory disease. It is unclear what forms of cellular and tissue mechanisms underly such a diverse set of non-cancer disorders.

2.3.1.4 The biological concept of the effective dose

The basic idea of the radiation protection concept of ICRP is the idea of a primary limiting quantity in radiological protection, the *effective dose* (see Chapter 5 of the present Volume, Chapter 4 of Landolt-Börnstein Vol. VIII/4 “Radiological Protection” [05Kau], and ICRP Publ. 103 [07ICR]). It is characterized by relating the risk of exposure to ionizing radiation from external or internal irradiation to a single dose quantity which takes account of the man as receptor, the different radiation sensitivities of various organs and tissues, and of the different radiation qualities. The different radiosensitivity of the various organs and tissues in the human body with respect to cancer induction and mortality is described by so-called *tissue weighting factors*. They have been developed from a reference population of equal number of both sexes including children and the unborn child (fetus).

Table 2.3.1.1 gives tissue weighting factors w_T recommended by the ICRP [07ICR] for the assessment of the effective dose, based on the relative radiation detriment.

Table 2.3.1.1. Recommended tissue weighting factors w_T .

Tissue	w_T	Σw_T
Bone Marrow (red), Colon, Lung, Stomach, Breast, Remainder Tissues*)	0.12	0.72
Gonades	0.08	0.08
Bladder, Oesophagus, Liver, Thyroid	0.04	0.16
Bone Surface, Brain, Salivary Glands, Skin	0.01	0.04

*) Remainder tissues: Adrenals, Extrathoracic (ET) region, Gall bladder, Heart, Kidneys, Lymphatic nodes, Muscle, Oral mucosa, Pancreas, Prostate/Uterus/Cervix, Small Intestine, Spleen, Thymus.

Table 2.3.1.2 gives *radiation weighting factors* w_R as a general set considered by the ICRP [07ICR] to be appropriate for application in radiological protection. The different radiation qualities are body-averaged values each representing a mean value for the relative biological effectiveness of all tissues of the body from exposure to ionizing radiations.

Table 2.3.1.2. Recommended radiation weighting factors w_R .

Radiation type	Radiation weighting factor w_R
Photons	1
Electrons and muons	1
Protons and charged pions	2
Alpha particles, fission fragments, heavy ions	20
Neutrons	a continuous function of neutron energy

2.3.1.5 Summary

Deterministic effects in tissues and organs are the result of the loss of substantial numbers of cells, so-called stem cells, thereby cutting off the supply of functional cells. The consequence can be a temporary or permanent loss of tissue function which may be life threatening. A characteristic of the dose-response relationship for deterministic effects of ionizing radiations is that they are avoidable below a dose threshold. Knowledge of dose thresholds has been derived from the tolerance doses observed in radiotherapy. They vary with the tissue – whereby the gonads, the bone marrow, the gastrointestinal cells and the lens of the eyes, are most sensitive tissues.

Whereas the development of *stochastic effects*, i.e. the induction of cancer by ionizing radiations is quite good understood and may be described by initiation of irreversible cellular damage by damage to DNA leading to gene mutations in single target cells in tissues, there is still a number of important questions that remain to be answered in the assessment of the risks of radiation-induced cancer in human populations. Very limited information is available at lower dose and lower dose rates that are important for radiation protection purposes. Consequently the risks have to be assessed from the populations exposed at high doses and dose rates by applying appropriate so-called dose and dose rate effectiveness factors.

Under the general aspect of precaution in radiological protection the assumption is made that the incidence of radiation-induced cancer and hereditary disease increases linearly with the dose, with no threshold. Protection standards therefore had to be set to a degree that any risk is limited to an acceptable level. The concept of radiological protection is limiting this acceptable risk by a *primary limiting quantity*, the *effective dose*, taking into account the biological effectiveness of the various kinds of ionizing radiations, and the different radiation sensitivities of the various body organs and tissues.

2.3.1.6 References for 2.3.1

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2.3.2 Biological Effects of Non-Ionizing Radiations

Radiation and Biological Effects Biological Effects

J.H. BERNHARDT

2.3.2.1 Introduction

Non-ionizing radiation (NIR) is a general term for both radiations and fields that form part of the electromagnetic spectrum having insufficient radiated energy to produce ionization in the medium through which it passes. Except for a narrow band of wavelengths, called visual radiations (light), levels of NIR normality encountered in our environment are unperceived by any of the human senses until the intensity is above a threshold level and is detected as heat or some other physical or biological stimulation.

Scientific studies have indicated that exposures above certain threshold levels can cause not only detectable biological effects but potentially adverse health consequences. Since NIRs are ubiquitous in our natural living and working environment, and the biological studies conducted so far contains many inconsistencies, it is essential that scientific research continues to determine if health effects occur at low exposure levels.

In the next sections the different types of non-ionizing radiations will be treated as follows, whereby the order of the electromagnetic field ranges is within the sequence of increasing frequency:

- 2.3.2.2 Ultrasound,
- 2.3.2.3 Static and slowly varying electric and magnetic fields,
- 2.3.2.4 Time varying electric and magnetic field of frequencies less than 100 kHz,
- 2.3.2.5 Electromagnetic fields of frequencies above 100 kHz,
- 2.3.2.6 Optical radiation and lasers.

The chapters will be treated – as far as possible – within a homogeneous structure:

- Biological effects,
- Risk assessment,
- Recommendations for health protection and safety guidelines,
- Summary.

2.3.2.2 Ultrasound

2.3.2.2.1 Introduction

This Section describes biological effects of ultrasound exposure caused by thermal and non-thermal mechanisms, furthermore safety recommendations and guidelines. For further reading the reviews of [98Bar, 01Nyb] are recommended. The sections below are mainly based on these reviews.

2.3.2.2.2 Biological effects

The effects of increased temperature on biological systems have been extensively reviewed [92NCR, 94Bar]. Although many studies have demonstrated that hyperthermia is a common teratogen, the important question to be answered relates to the duration and degree of exposure required to produce the effect.

Animal studies reporting effects of whole-body temperature elevations of approximately 2°C typically involved exposure to hot air for periods of 60 min because it takes about 30 min to overcome the normal maternal homeostatic response and elevate the maternal core temperature.

Hyperthermia is a proven teratogen in mammalian biological systems. Many of the abnormalities reported in heat-exposed animals have also been found in children following *in utero* febrile episodes; therefore, heat exposure is considered to be a human teratogen [95Edw]. Smith and co-workers reported that maternal febrile illness that caused the human body temperature to rise above 38°C in early stages of pregnancy was associated with fetal anomalies [78Smi]. Long-term hyperthermia above 39°C may therefore be teratogenic to the human fetus. Data from retrospective studies indicate that mothers of babies with various CNS malformations experienced increased prevalence of febrile illness during early pregnancy.

Thermal Effects

The amount of ultrasound induced temperature increase occurring during an ultrasonic exposure depends on the properties of both the ultrasound field parameters and the biological tissue involving ultrasound absorption, thermal conduction and blood perfusion. For water and body liquids (urine, amniotic fluid, aqueous humour of the eye, cerebrospinal fluid, endolymph of the inner ear) there is little absorption in acoustic conditions of linear propagation, and therefore small risk of unwanted temperature increase.

Absorption of ultrasound in biological tissues strongly depends on the molecular composition of the constituent tissue. The absorption coefficient increases as a function of protein content, with collagen having a particularly high specific absorption [92NCR]. The acoustic properties of biological tissue are comprehensively reviewed in a book by Duck [90Duc]. Higher absorption values, with the potential for undesirable ultrasound induced temperature increase, are found in skin, tendon and spinal cord. The highest value is obtained in mineralised bone.

The extent of ultrasound-induced tissue heating depends on the balance of heat gain and heat loss. The tissue absorption characteristics determine how much heat is deposited, whereas its composition and extent of vascularity determine how much heat is lost from the system by conduction and perfusion. The final tissue temperature achieved is dependent on thermal conduction into neighbouring tissues and blood perfusion dissipating heat, especially after protracted exposure. In internal organs such as the liver, kidney and brain, the perfusion rate is permanently high. Perfusion is poor in fatty tissue, tendon, sclera, periosteum and bone. Perfusion of skin and muscles (including heart and bowels) changes depending on temperature and activity. Unperfused tissues include the cornea, the lens and the vitreous body of the eye. Hence, the human lens may be susceptible to significant levels of ultrasound-induced heating.

The periosteum is an irritable structure, which maybe expected to sense, as pain, an unphysiological temperature increase. The fetus is, however, unable to elicit such a response in the mother, and therefore fetal bone is of special concern. During prenatal life, bone mineralization starts in the twelfth week of pregnancy and continues through birth to early childhood. There is a relationship between the amount of ultrasound-induced heating and gestational age that correlates with bone development; fetal bone structures become increasingly prone to heating with advancing pregnancy. In late pregnancy the extent of tissue damage is small, relative to the size of the fetus, and associated bioeffects may be difficult to detect. In embryonic stages of development, any adverse physical insult can produce bioeffects that are easier to detect, since they are likely to be more catastrophic, but the likelihood of producing a significant temperature increase in soft embryonic tissue is lower than in bony tissue of the fetus.

As collagen can have the highest acoustic absorption coefficients, the tissues that are likely to be heated to the greatest extent are bone and collagen and nearby soft tissue. Thus, the fetal cerebral cortex can be significantly heated by pulsed ultrasound at diagnostic intensities, particularly in late gestation [93Bos]. Greater temperature increases occur when bone is situated within the ultrasound beam. Temperature increases of approximately 5°C have been reported at the brain (bone tissue interface close to the inner aspect of the skull parietal bone in guinea-pig fetuses [93Bos]. Similar results (temperature increase of 5.1°C) were recorded using the same exposure conditions in dead guinea-pig fetuses [98Hor].

The neurones of the developing fetal brain are considered to be the most susceptible to the effects of ultrasound-induced heating [92WFU]. Also the tissues of the central nervous system (CNS) have a relatively low absorption coefficient; they are encased in the bone of the fetal skull and vertebrae, which are susceptible to significant heating by ultrasound. Soft tissue adjacent to the cerebral cortex is heated by

conduction. Exposure to some diagnostic equipment operating in pulsed Doppler mode has been shown to produce biologically-significant temperature increases in tissue, particularly when bone is present. The biological consequences of a hyperthermic episode depend on the magnitude of temperature elevation and the duration of exposure. Data are available for whole-body exposures (generally resulting in severe abnormalities) where it has been shown that rat embryos exposed to a temperature increase of 4°C for 5 min developed encephaloceles [93Edw].

There is some evidence to suggest that ultrasound-induced bioeffects can be potentiated by modest increases in temperature. In general, febrile patients may be considered to be at a greater risk from hyperthermic damage, since their elevated core temperature would add to the ultrasound-induced heating of the embryo or fetus [92WFU]. Hyperthermia is generally accepted as a human teratogen [92WFU].

Cavitation effects

At moderate acoustic pressure amplitudes, a precondition for inertial (previously termed “transient”) cavitation is the existence of gas bodies, i.e., gaseous inclusions. Gas inclusions are normal in the respiratory tract and in the bowels. At high peak pressures, inertial cavitation is capable of causing cavities in liquids to collapse at speeds that exceed the speed of sound in the gas. This results in the production of sufficient energy to disrupt chemical bonds and chemically reactive free radicals. These free radicals, theoretically, may interfere with DNA, causing chromosomal damage, although this has never been observed in either patients or animals. This mechanism of rapid bubble growth and collapse can continue throughout the exposure period. In contrast, noninertial (stable) cavitation involves the continued pulsation of pre-existing gas-filled bubble, without collapse, and with capabilities for exerting mechanical stress without accompanying chemical action involving free radicals. Body fluids saturated with gas molecules and protein nuclei might be expected to support cavitation by providing the sites for bubble formation during the negative pressure phase of the acoustic wave.

Evidence of inertial cavitation occurring in mammalian tissues is based on data from lithotripter exposures. Cavitation-related bioeffects have been reported at gas/tissue interfaces, such as in the mammalian lung, at low-level (equivalent to diagnostic) exposures with pulsed ultrasound [94Tar]. The effects are described as extravasation, or pulmonary capillary bleeding. The exact mechanism responsible for the cavitation-related effect is yet to be identified, but it is not the thermally induced. It cannot be assumed that cavitation nuclei and gas bubbles do not exist in human tissue. They may be expected to be present under certain clinical conditions such as surgical interventions, traumatic lesions with lesions of skin, lungs or bowels, infusions, decompression illness and clostridial infection.

Ultrasound contrast agents can increase the likelihood of occurrence and the extent of ultrasound induced bioeffects *in vivo*. Effects reported in animal studies at diagnostically relevant pulsing conditions include haemolysis and microvascular rupture in muscle and kidney. Extrapolation of ultrasound exposure, contrast agent dose and physiological conditions from small animals to humans has not been validated [07Bar]. The clinical significance of these findings requires further investigation.

The presence of ultrasound contrast agent in the blood increases the vulnerability of many mammalian tissues and organs to damage from exposure to lithotripter fields *in vivo*. The increased vulnerability to exposure can persist for hours after the initial injection of contrast agent [07Bar].

Concerning the clinical relevance of tissue lesions it is referred to the literature (i.e., [97Bar, 01Nyb]).

2.3.2.2.3 Recommendations for health protection and safety guidelines

Since the safety aspects of ultrasound in medicine are treated in detail in Sect. 5.4.2 only a short summary is presented here.

International recommendations and guidelines

Modern sophisticated ultrasonic equipment is capable of delivering substantial levels of acoustic energy into the body when used at maximum output. The risk of producing bioeffects has been studied by international expert groups during symposia supported by the World Federation for Ultrasound in Medicine and Biology (WFUMB). These symposia have resulted in the publication of internationally accepted conclusions and recommendations. National ultrasound safety committees have published

guidelines as well. These recommendations and safety guidelines offer valuable information to help users apply diagnostic ultrasound in a safe and effective manner. However, there is also a modern trend towards self-regulation which has implications for the worldwide use of diagnostic ultrasound. It has resulted in a move away from the relatively simple scheme of FDA-enforced, application-specific limits on acoustic output to a scheme whereby risk of adverse ultrasound exposure is assessed from information provided by the equipment in the form of a real-time display of safety indices. The shift of responsibility for risk assessment from a regulatory authority to the user creates an urgent need for awareness of risk and the development of knowledgeable and responsible attitude to safety issues. To encourage this approach, it is incumbent on authorities, ultrasound societies and expert groups to provide relevant information on biological effects that might result from ultrasonic procedures. It is obvious from the continuous stream of inquiries received by ultrasound societies that effective dissemination of such knowledge requires sustained strenuous effort on the part of ultrasound safety committees. There is a strong need for continuing education to ensure that appropriate risk/benefit assessments are made by users based on an appropriate knowledge of the probability of biological effects occurring with each type of ultrasound procedure. Concerning the current safety guidelines and international recommendations the actual versions can be downloaded from www.wfumb.org, www.aium.org, www.efsumb.org.

2.3.2.2.4 Summary

Mammalian tissues have differing sensitivities to damage by physical agents such as ultrasound. Actively dividing cells of the embryonic and fetal central nervous system are most readily disturbed. As a diagnostic ultrasound beam envelopes a small volume of tissue, it is possible that the effects of mild disturbance may not be detected unless major neural pathways are involved. There is evidence that ultrasound can be detected by the central nervous system; however, this does not necessarily imply that the bioeffect is hazardous to the fetus. Biologically significant temperature increases can occur at or near to bone in the fetus from the second trimester, if the beam is held stationary for more than 30 s in some pulsed Doppler applications. It is important in these applications to minimize the acoustic output and duration of exposure. Reports in animals and humans of retarded growth and development following frequent exposures to diagnostic ultrasound, in the absence of significant heating, are difficult to explain from the current knowledge of ultrasound mechanisms. Thresholds for teratogenic effects of hyperthermia are determined by a combination of the induced temperature elevation and its duration. Most ultrasound diagnostic procedures involve brief dwell times, which help to widen the safety margin. A diagnostic exposure that produces a maximum temperature rise of 1.5°C above normal physiological levels (37°C) does not present a risk from thermal effects in humans, regardless of the duration of exposure.

There is no evidence of cavitation effects occurring in the soft tissue of the fetus when exposed to diagnostic ultrasound; however, the possibility exists that such effects may be enhanced by the introduction of echo-contrast agents.

Safety guidelines

The regulatory process that controls acoustic output from ultrasound medical devices has been largely dictated by the Food and Drug Administration (FDA) in the USA. In recent years there have been significant changes that make the development of an internationally acceptable standard on ultrasound safety both complicated and imperative. The trend towards self-regulation has resulted in a move away from the relatively simple, but outdated, scheme of enforced application-specific limits on acoustic output. The FDA now provides an option for manufacturers to obtain market approval for medical ultrasound devices that can increase the intensity at the fetus by almost a factor of 8, provided that an output display is incorporated into the equipment design.

The combination of recent changes in the regulation of medical ultrasound and continuing development in international standards will allow greater control by the ultrasonologist of higher output power levels used in diagnostic equipment. Major national ultrasound societies have formulated guidelines for the safe use of ultrasound in medicine. There is a strong need for continuing education to ensure that appropriate risk/benefit assessments are made by ultrasonologists based on an understanding of the probability of biological effects occurring with each type of ultrasound procedure. The relaxation

of intensity limits for pre-market approval by the FDA allows substantially increased intensity of ultrasound to be delivered to the fetus. Modern diagnostic equipment operating at maximum output conditions can produce significant biological effects in mammalian tissues. These issues have been addressed by the World Federation for Ultrasound in Medicine and Biology in published conclusions and recommendations that represent international consensus on safe use of ultrasound in medicine.

2.3.2.3 Static and slowly varying electric and magnetic fields

2.3.2.3.1 Biological effects: experimental and epidemiological data

In vitro studies, animal studies, and studies of effects in humans either in the laboratory or through epidemiological investigation are described separately [06WHO].

In vitro experiments

The results of in vitro studies are useful for elucidating interaction mechanisms, and for indicating the sorts of effects that might be investigated *in vivo*. However, they are not sufficient to identify health effects without corroborating evidence from *in vivo* studies.

No in vitro experiments have been described with exposure to static electric fields.

A number of different biological effects of static magnetic fields have been explored *in vitro* [05Miy]. Different levels of biological organization at a cellular level have been investigated, including cell free systems (isolated membranes, enzymes or biochemical reactions) and various cell models (using both bacteria and mammalian cells). Endpoints studied included cell orientation, cell metabolic activity, cell membrane physiology, gene expression, cell growth, and genotoxicity.

Positive and negative findings have been reported for all these endpoints. However, most data have not been replicated. The observed effects are rather diverse and were found after exposure to a wide range of magnetic flux densities. There is evidence that static magnetic fields can affect several endpoints at intensities lower than 1 T, mainly in the mT range. Thresholds for some of the effects were reported, but other studies indicated non-linear responses without clear threshold values.

Overall, the *in vitro* data do not present a clear picture and they are not indicative of any health effects.

Animal studies

Few animal studies on the effects of static electric fields have been carried out. No effects have been noted other than those associated with the perception of the surface electric charge.

A large number of animal studies on the effects of static magnetic fields have been carried out. The most consistent responses seen in neurobehavioral studies suggest that the movement of laboratory rodents in static magnetic fields equal to or greater than 4 T may be unpleasant, inducing aversive responses and conditioned avoidance [03Hou].

There is good evidence that exposure to fields greater than about 1 T (0.1 T in larger animals) will induce flow potentials around the heart and major blood vessels, but the physiological consequences of this remain unclear [05Ten] Several hours of exposure to very high flux densities of up to 8 T in the heart region did not result in any cardiovascular effects in pigs [99Kan]. In rabbits, short and long exposures to fields, ranging from geomagnetic levels to the millitesla range have been reported to affect the cardiovascular system, although the evidence is not strong.

Reproduction and development are very important issues in MRI exposure of both patients and clinical staff. In this respect, only a few good studies of static magnetic fields are available at field values above 1 T. MRI studies are uninformative because the effect of the static field cannot be distinguished from the possible general effects of the radiofrequency and pulsed gradient fields. Further examination is urgently needed to assess the health risk. In general, so few animal studies have been carried out with regard to genotoxicity and cancer that it is not possible to draw any firm conclusions.

Human studies

Static electric fields do not penetrate electrically conductive objects such as the human body; the field induces a surface electric charge. The perception threshold for the effects of surface electric charge in people depends on various factors and can range between $10\text{--}45 \text{ kV m}^{-1}$. Annoying sensation thresholds are probably equally variable but have not been systematically studied.

Endpoints investigated in human experimental studies of static magnetic field exposure have included peripheral nerve function, brain activity, neurobehavioral and cognitive function, sensory perception, cardiac function, blood pressure, heart rate, serum proteins and hormone levels, and body and skin temperature. The results indicate that there are no effects of static magnetic field exposure at up to 8 T on neurophysiologic responses and cognitive functions in stationary volunteers [03Cha1]. A dose-dependent induction of vertigo and nausea was found in workers, patients, and volunteers during movement in static fields greater than about 2 T [92Sch]. One study suggested that hand-eye coordination and near visual contrast sensitivity is reduced in fields adjacent to an 1.5 T MRI unit [03DeV]. Occurrence of these effects is likely to be dependent on the gradient of the field and the movement of the subject. A small change in blood pressure and heart rate was observed in volunteers exposed to an 8 T static field [03Cha2], but were in the range of normal physiological variability. There is no evidence of effects of static magnetic fields on other aspects of cardiovascular physiology or on serum proteins and hormones. Exposure did not appear to induce temperature changes in humans.

It was concluded that most of the studies were very small, based on convenience samples, and often included non-comparable groups. Thus, it was not possible to draw any conclusions regarding the wide variety of endpoints examined.

Epidemiological studies

Epidemiological studies have been carried out almost exclusively on workers exposed to static magnetic fields generated by equipment using large DC currents (reviewed by [05Fey]). Most workers were exposed to moderate static magnetic fields of up to several tens of mT, either as welders, Aluminium smelters, or workers in various industrial plants using large electrolytic cells in chemical separation processes. However, such work is also likely to have involved exposure to a variety of potentially hazardous fumes and aerosols, thus confounding interpretation. Health endpoints studied in these workers include cancer, haematological changes and related outcomes, chromosome aberration frequency, reproductive outcomes, and musculoskeletal disorders. In addition, a few studies examined fertility and pregnancy outcome in female MRI operators, where the potential to have been exposed to relatively large static fields of up to 1 T may have existed. Two studies examined pregnancy outcome in healthy volunteers exposed to MRI examinations during pregnancy.

Increased risks of various cancers, e.g., lung cancer, pancreatic cancer, and haematological malignancies, were reported, but results were not consistent across studies. The few epidemiological studies published to date leave a number of unresolved issues concerning the possibility of increased cancer risk from exposure to static magnetic fields. Assessment of exposure has been poor and the number of participants in some of the studies has been very small, so these studies are able to detect only very large risks for such rare diseases. The inability of these studies to provide useful information is confirmed by the lack of clear evidence for other, more established carcinogenic factors present in some of the work environments. Other non-cancerous health effects have been considered even more sporadically. Most of these studies, including those concerning reproduction, are based on very small numbers and have numerous methodological limitations.

At present, there are inadequate data for a health evaluation of these long-term effects, and it is difficult to draw any conclusions. It was considered that environments with a potential for high field exposure, such as those associated with MRI have not been adequately evaluated.

2.3.2.3.2 Health risk assessment and exposure guidelines

The overall analysis of health risks from static electric and magnetic fields in the WHO EHC is as follows [06WHO, 07Van].

Static electric fields

There are no studies on exposure to static electric fields from which any conclusions on chronic or delayed effects can be made. The International Agency for Research on Cancer (IARC) noted there was insufficient evidence to determine the carcinogenicity of static electric fields [02IAR]. Few studies of the acute effects of static electric fields have been carried out. On the whole, the results suggest that the only adverse acute health effects are associated with direct perception of fields and discomfort from microshocks.

Static magnetic fields

The available evidence from epidemiological and laboratory studies are insufficient to draw any conclusions with regard to chronic and delayed effects. IARC (2002, [02IAR]) concluded that there was inadequate evidence in humans for the carcinogenicity of static magnetic fields, and no relevant data available from experimental animals. The carcinogenicity of static magnetic fields to humans is therefore not at present classifiable.

Short-term exposure to static magnetic fields in the Tesla range and associated field gradients induce a number of acute effects. Cardiovascular responses, such as changes in blood pressure and heart rate, have been occasionally observed in human volunteer and animal studies. However, these were within the range of normal physiology for exposure to static magnetic fields up to 8 T. Although not experimentally verified, it is important to note that calculations suggest minor changes in heartbeat, the induction of ectopic heartbeats, and an increase in the likelihood of re-entrant arrhythmia, possibly leading to ventricular fibrillation. The first two effects are thought to have thresholds in excess of 8 T, and threshold values for the third are difficult to assess at present. The limitations of the available data are such, however, that it is not possible to draw firm conclusions about the effects of static magnetic fields on the endpoints considered above.

Physical movement within a static field gradient has been reported to induce sensations of vertigo and nausea, and sometimes phosphenes and a metallic taste in the mouth, for static fields in excess of about 2-4 T. Although only transient, such effects may adversely affect people. Together with possible effects on hand-eye coordination, the optimal performance of workers executing delicate procedures could be reduced, with a concomitant impact on safety.

Effects on other physiological responses have been reported, but it is difficult to reach any firm conclusion without independent replication or confirmation.

ICNIRP static field guidelines; 1994 and 2009

The ICNIRP (1994) guidelines [94ICN] were formulated at a time when MRI was a relatively new procedure, and there was very limited knowledge of the effects of exposure to static magnetic fields much in excess of 2-4 T. These guidelines are now being re-assessed by the ICNIRP in the light of the WHO EHC health risk analysis described above. The revised guidelines are published in 2009 [09ICN1].

There are no reports of long term or chronic adverse effects following prolonged static magnetic field exposure, but few data are available on which to base any judgment.

Electric fields

No exposure limits were given since only indirect effects (spark discharges) have been described.

Magnetic fields

Based on review of the scientific evidence, ICNIRP recommends the following limits for exposure [09ICN1]:

Occupational exposures

It is recommended that occupational exposure of the head and trunk should not exceed a spatial peak magnetic flux density of 2 T. However, for specific work applications, exposure up to 8 T can be permitted, if the environment is controlled and appropriate work practices are implemented to control movement-induced effects. Sensory effects due to movement in the field can be avoided by complying with basic restrictions set in the ELF guidelines. When restricted to the limbs, maximum exposures of up to 8 T are acceptable.

General public exposures

Acute exposure of the general public should not exceed 400 mT (any part of the body), reflecting a reduction factor of 5 with respect to the occupational limits. However, because of potential indirect adverse effects, ICNIRP recognizes that practical policies need to be implemented to prevent inadvertent harmful exposure of people with implanted electronic medical devices and implants containing ferromagnetic materials, and injuries due to flying ferromagnetic objects, and these considerations can lead to much lower restriction levels, such as 0.5 mT [02IEC]. The exposure limits to be set with regard to these non-biological effects are not, however, the remit of ICNIRP.

The rationale for these guidelines limits can be found in full in [09ICN1].

[Table 2.3.2.1](#) summarises the limits.

Table 2.3.2.1. Limits of exposure^{a)} to static magnetic fields. From [09ICN1].

Exposure characteristics	Magnetic flux density
Occupational^{b)}	
Exposure of head and of trunk	2 T
Exposure of limbs ^{c)}	8 T
General public^{d)}	
Exposure of any part of the body	400 mT

^{a)} ICNIRP recommends that these limits should be viewed operationally as spatial peak exposure limits.

^{b)} For specific work applications, exposure up to 8 T can be justified, if the environment is controlled and appropriate work practices are implemented to control movement-induced effects.

^{c)} Not enough information is available on which to base exposure limits above 8 T.

^{d)} Because of potential indirect adverse effects, ICNIRP recognizes that practical policies need to be implemented to prevent inadvertent harmful exposure of persons with implanted electronic medical devices and implants containing ferromagnetic material, and dangers from flying objects, which can lead to much lower restriction levels such as 0.5 mT.

2.3.2.3.3 Summary

Few animal studies on the effects of static electric fields have been carried out. No effects have been noted, other than those associated with the perception of the surface electric charge.

A large number of animal studies on the effects of static magnetic fields have been carried out. The most consistent responses seen in neurobehavioral studies suggest that the movement of laboratory rodents in static magnetic fields equal to or greater than 4 T may be unpleasant, inducing aversive responses and conditioned avoidance.

Static electric fields do not penetrate electrically conductive objects such as the human body; the field induces a surface electric charge. The perception threshold for the effects of surface electric charge in people depends on various factors and can range between 10–45 kV m⁻¹. Annoying sensation thresholds are probably equally variable but have not been systematically studied.

Endpoints investigated in human experimental studies of static magnetic field exposure have included peripheral nerve function, brain activity, neurobehavioral and cognitive function, sensory perception, cardiac function, blood pressure, heart rate, serum proteins and hormone levels, and body and skin temperature. The results indicate that there are no effects of static magnetic field exposure at up to 8 T on neurophysiologic responses and cognitive functions in stationary volunteers. A dose-dependent induction of vertigo and nausea was found in workers, patients, and volunteers during movement in static fields greater than about 2 T.

The main conclusions are that no acute effects other than transient phenomena such as vertigo and nausea have been observed with exposure to static magnetic flux densities up to 8 T. There are no reports of long term or chronic adverse effects following prolonged static magnetic field exposure, but few data are available on which to base any judgment. The guidelines on static field exposure published by ICNIRP in 2009 recommends that occupational exposure of the head and trunk should not exceed a

spatial peak magnetic flux density of 2 T. However, for specific work applications, exposure up to 8 T can be permitted, if the environment is controlled and appropriate work practices are implemented to control movement-induced effects. Acute exposure of the general public should not exceed 400 mT (any part of the body), reflecting a reduction factor of 5 with respect to the occupational limits. However, because of potential indirect adverse effects, ICNIRP recognizes that practical policies need to be implemented to prevent inadvertent harmful exposure of people with implanted electronic medical devices and implants containing ferromagnetic materials, and injuries due to flying ferromagnetic objects, and these considerations can lead to much lower restriction levels, such as 0.5 mT.

2.3.2.4 Time varying electric and magnetic fields of frequencies less than 100 kHz

2.3.2.4.1 Introduction

Many effects on biological systems exposed to ELF fields have been reported. Explicit distinctions must be made between the concepts of interaction, biological effect, and health hazard, consistent with the criteria used by international bodies when making health assessments [02ICN]: Biological effects occur when fields interact to produce responses that may or may not be experienced by people. Deciding whether biological changes have health consequences depends, in part, on whether they are reversible, are within the range for which the body has effective compensation mechanisms, or are likely, taking into account the variability of response among individuals, to lead to unfavourable changes in health.

While it is known that electromagnetic interference with electromedical devices leads to adverse health consequences, this review is only concerned with direct effects of the fields on living tissues.

For further reading the following reviews are mentioned: [92Ber, 93WHO, 01AGN, 03ICN, 07WHO].

2.3.2.4.2 Biological effects

Neurobehaviour

Neurobehavioural studies encompass the effects of exposure to ELF electromagnetic fields on the nervous system and its responses at different levels of organization. These include the direct stimulation of peripheral and central nerve tissue, perceptual effects resulting from sensory stimulation, and effects on central nervous system function. Effects on the latter can be assessed both electrophysiological by recording the electrical activity of the brain, and by tests of cognition, assessment of mood, and other studies.

The nervous system also has a central role in the control of other body systems, particularly the cardiovascular system, through direct nervous control, and the endocrine system, through neural input into the pineal and pituitary glands. These glands in turn influence reproduction and development, and in a more general way, physiology and well-being.

The brain and nervous systems function by using electrical signals, and may therefore be considered particularly vulnerable to low frequency EMF's and the resultant induced electric fields and currents. Substantial numbers of laboratory experiments with volunteers and animals have investigated the possible consequences of exposure to weak EMF's on various aspects of nervous system function, including cognitive, behavioural and neuroendocrine responses.

These studies have been reviewed by [97NRC, 98NIE, 02IAR, 03ICN, 04McK].

Electrophysiological considerations

An examination of the electrophysiological properties of the nervous system, particularly the central nervous system (CNS, brain and spinal cord) gives an indication of its likely susceptibility to the electric fields induced in the body by EMF exposure. Ion channels in cell membranes allow passage of particular ionic species across the cell membrane in response to the opening of a "gate" which is sensitive to the transmembrane voltage [03Mat]. It is well established that electric fields induced in the body either by

direct contact with external electrodes, or by exposure to low frequency magnetic fields, will, if of sufficient magnitude, excite nerve tissue through their interaction with these voltage-gated ion channels. Sensitivity is therefore primarily to the transmembrane electric field and varies widely between different ion channels [02Sau]. Many voltage-gated ion channels are associated with electrical excitability and electrical signalling. Such electrically excitable cells not only comprise neurons, glial and muscle cells, but also endocrine cells of the anterior pituitary, adrenal medulla and pancreas, gametes and, with reservations, endothelial cells [01Hil].

Peripheral nerves comprise neurons whose cell bodies are located within the CNS with extended processes (axons) that lie outside the CNS. They conduct action potentials (impulses) towards (sensory nerves) or from (motor nerves) the spinal cord and nerve stimulation shows an all-or-nothing threshold behaviour. Excitation results from a membrane depolarisation between 10-20 mV, corresponding to an electric field in tissue of 5-25 V m⁻¹ [04McK]. Pulsed magnetic fields, where the rate of change of field induces large localised electric fields, can directly stimulate peripheral nerves and nerve fibres located within the brain (see below).

Cells of the central nervous system are considered to be sensitive to electric fields induced in the body by exposure to ELF magnetic fields at levels that are below threshold for impulse initiation in nerve axons. Such weak electric field interactions have been shown in experimental studies mostly using isolated animal brain tissue to have physiological relevance. Jefferys and colleagues identified in vitro electric field thresholds of around 4-5 V m⁻¹ [03Jef].

The CNS *in vivo* is likely to be more sensitive to induced low frequency electric fields and currents than are in vitro preparations [02Sau]. Much of normal cognitive function of the brain depends on the collective activity of very large numbers of neurons; neural networks are thought to have complex non-linear dynamics that can be very sensitive to small voltages applied diffusely across the elements of the network [03ICN, 03Jef]. Experimental work confirmed a neural network threshold of around 140 mV m⁻¹ which the authors found was lower than single neuron thresholds [03Fra].

Surface electric charge

An electric charge is induced on the surface of a human (or other living organism) exposed to a low frequency electric field that alternates in amplitude with the frequency of the applied field. The surface electric charge can be perceived directly through the induced vibration of body hair and tingling sensations in areas of the body, particularly the arms, in contact with clothing, and indirectly through spark discharges between a person and a conducting object within the field. The threshold for direct perception has shown wide individual variation; 10% of the exposed subjects had detection thresholds of around 2-5 kV m⁻¹ at 50 or 60 Hz, whereas 50% could detect fields of 7-20 kV m⁻¹ [88Ber, 98Rei]. These effects were considered annoying by 50% of the test subjects exposed under laboratory conditions above electric field strengths of about 15-20 kV m⁻¹. In addition to showing a wide variation in individual sensitivity, these responses also vary with environmental conditions, particularly humidity.

It has been estimated that spark discharges would be painful to 70% of subjects who are well-insulated and who touch a grounded object within a 5 kV m⁻¹ field whereas they would be painful to about 50% in a 10 kV m⁻¹ field [88Ber, 98Rei]. Unpleasant spark discharges can also occur when a grounded person touches a large conductive object such as a large vehicle that is “well-insulated” from ground and is situated within a strong electric field. Here, the threshold field strength required to induce such an effect varies inversely with the size of the conductive object.

People can perceive electric currents directly applied to the body through touching, for example, a conductive loop in which current is induced by exposure to environmental electromagnetic fields. Thresholds for directly applied currents have also been characterised. At 50 to 60 Hz, the male median threshold for perception was between 0.36 mA (finger contact) and 1.1 mA (grip contact), while pain occurred at 1.8 mA (finger contact). Median thresholds for women were generally found to be two thirds of the male thresholds, while children were assumed to have median thresholds half of male threshold values [93WHO]. There is also a wide variety in the individual's ability to detect currents, there is, for example, about one order of magnitude difference in the perception threshold at the 0.5 percentile and the 99.5 percentile at 50/60 Hz. Generally, the ability to detect fields or currents decreases with increasing frequency. This has been characterised for the perception of currents, the threshold is increasing by about

two orders of magnitude at higher frequencies: 0.36 mA at 50/60 Hz, 4 mA at 10 kHz and 40 mA at 100 kHz [93WHO].

Nerve stimulation

Large, rapidly changing, pulsed magnetic fields used in various specialised medical applications such as magnetic resonance imaging (MRI) and transcranial magnetic stimulation (TMS) can induce electric fields large enough to stimulate nervous tissue in humans. Minimum, orientation-dependent stimulus thresholds for large diameter ($20\text{ }\mu\text{m}$) myelinated nerve axons have been estimated to be approximately 6 V m^{-1} at frequencies up to about 1-3 kHz [98Rei, 99Rei]. In MRI, nerve stimulation is an unwanted side effect of a procedure used to derive cross-sectional images of the body for clinical diagnosis. Threshold rates of change of the switched gradient magnetic fields used in MRI for perception, discomfort and pain resulting from peripheral nerve stimulation are extensively reviewed by Nyenhuis et al. [01Nye]. Generally, median, minimum threshold rates of change of magnetic field (during periods of $< 1\text{ ms}$) for perception were $15\text{-}25\text{ }\mu\text{T s}^{-1}$ depending on orientation and showed considerable individual variation. Thresholds rose as the pulse width of the current induced by the switched gradient field decreased, the median pulse width (the chronaxie) corresponding to a doubling of the minimum threshold (the rheobase) ranged between 360 and 380 μs but again showing considerable individual variation [99Bou].

In TMS, parts of the brain are deliberately stimulated in order to produce a transient, functional impairment for use in the study of cognitive processes [98Rei, 99Uen]. Furthermore, in TMS, brief, localised, suprathreshold stimuli are given, typically by discharging a capacitor through a coil situated over the surface of the head, in order to stimulate neurons in a small volume (a few cubic centimeters) of underlying cortical tissue. The induced current causes the neurons within that volume to depolarise synchronously, followed by a period of inhibition [00Fit]. Reilly [98Rei] noted induced electric field thresholds to be of the order of 20 V m^{-1} . However, Walsh et al. cited typical rates of change of magnetic field of 30 kT s^{-1} over a $100\text{ }\mu\text{s}$ period transiently inducing an electric field of 500 V m^{-1} in brain tissue [98Wal].

Retinal function

The effects of exposure to weak low frequency magnetic fields on human retinal function are well established. Exposure of the head to magnetic flux densities above about 5 mT at 20 Hz , rising to about 15 mT at 50 Hz , will reliably induce faint flickering visual sensations called magnetic phosphenes [03Att]. It is generally agreed that these phosphenes result from the interaction of the induced electric current with electrically sensitive cells in the retina.

There is good reason to view retinal circuitry as an appropriate model for induced electric field effects on CNS neuronal circuitry in general [03Att]. Firstly, the retina displays all the processes present in other CNS areas, such as graded voltage signalling and action potentials, and have a similar biochemistry. Secondly, in contrast to more subtle cognitive effects, phosphenes represent a direct and reproducible perception of field interaction. A clear distinction can be made in this context between the detection of a normal visual stimulus and the abnormal induction of a visual signal by non-visual means [03Sau]; the latter suggests the possibility of direct effects on cognitive processes elsewhere in the CNS.

Thresholds for electrically induced phosphenes have been estimated to be about $10\text{-}14\text{ mA m}^{-2}$ at 20 Hz [77Adr]. A similar value (10 mA m^{-2} at 20 Hz), based on studies of magnetically induced phosphenes, has been derived by Wake et al. [98Wak]. The equivalent electric field threshold can be estimated as around $100\text{-}140\text{ mV m}^{-1}$ using a tissue conductivity for brain tissue of about 0.1 S m^{-1} .

However, Taki et al. indicated that calculations of phosphene thresholds suggested that electrophosphene thresholds were around 100 mV m^{-1} , whereas magnetophosphene thresholds were around 10 mV m^{-1} at 20 Hz [03Tak].

Animal studies

Various animal models have been used to investigate possible field induced effects on brain function and behaviour. These include effects on neurotransmitter levels, electrical activity, field detection and the performance of learned tasks. Overall, a few field-dependent responses have been identified but even the most consistent effects appear small in magnitude and transient in nature.

2.3.2.4.3 Other effects

In addition to the surface charging effects and electrophysiological responses described above, the possibility that exposure to low frequency EMFs may induce other biological effects has continued to be investigated. Many different models, including volunteers, animals and cultures of cells have been examined using a wide range of tests and exposure conditions. These studies have been reviewed by [03ICN, 04McK, 07WHO].

Neuroendocrine system

The results of volunteer studies as well as residential and occupational studies suggest that the neuroendocrine system is not adversely affected by exposure to power-frequency electric and/or magnetic fields [07WHO]. This applies particularly to the circulating levels of specific hormones of the neuroendocrine system, including melatonin, released by the pineal gland, and a number of hormones involved in the control of body metabolism and physiology, released by the pituitary gland.

Neurodegenerative disorders

It has been hypothesized that exposure to ELF fields is associated with several neurodegenerative diseases. For Parkinson disease and multiple sclerosis the number of studies has been small and there is no evidence for an association with these diseases. For Alzheimer disease and amyotrophic lateral sclerosis (ALS) more studies have been published. Some of these reports suggest that people employed in electrical occupations have an increased risk of ALS. So far no biological mechanism has been established which can explain this association, although it could have arisen because of confounders related to electrical occupations such as electric shocks. Overall, the evidence for the association between ELF exposure and ALS is considered inadequate.

Cardiovascular disorders

Experimental studies of both short- and long-term exposure indicate that, while electric shock is an obvious health hazard, other hazardous cardiovascular effects associated with ELF fields are unlikely to occur at exposure levels commonly encountered environmentally or occupationally. Although various cardiovascular changes have been reported in the literature, the majority of effects are small and the results have not been consistent within and between studies. Overall, the evidence does not support an association between ELF exposure and cardiovascular disease.

Immune system and haematology

Evidence for the effects of ELF electric or magnetic fields on components of the immune system is generally inconsistent. Many of the cell populations and functional markers were unaffected by exposure.

There have been few studies carried out on the effects of ELF magnetic fields on the haematological system. In experiments evaluating differential white blood cell counts, exposures range from 2 µT to 2 mT. No consistent effects of acute exposure to magnetic fields or to combined electric and magnetic fields have been found in either human or animal studies.

Overall, the evidence for effects of ELF electric or magnetic fields on the immune system and haematological system is considered inadequate [07WHO].

Reproduction and development

Epidemiological studies have not shown an association between adverse human reproductive outcomes and maternal or paternal exposure to ELF fields. The evidence for developmental effects and for reproductive effects is inadequate [07WHO].

2.3.2.4.4 Cancer

The possibility that exposure to low frequency EMF's increases the risk of cancer has been subject to much epidemiological and experimental research over the last two decades and has been widely reviewed

by national and international expert groups (e.g. [01AGN, 02IAR, 03ICN, 98NIE]. The association between childhood leukaemia and residential ELF magnetic fields, first identified by Wertheimer & Leeper [79Wer] and subsequently found in a number of epidemiological studies, has driven experimental and epidemiological research and risk assessment forwards in this area and led to the classification of ELF magnetic fields by the International Agency for Research on Cancer (IARC) as a “possible human carcinogen” [02AR1].

Since the first report suggesting an association between residential ELF magnetic fields and childhood leukaemia was published in 1979, dozens of increasingly sophisticated epidemiological studies have examined this association. In addition, there have been numerous comprehensive reviews, meta-analyses, and two pooled analyses. In one pooled analysis based on nine well-conducted studies, virtually no association was noted for exposure to ELF magnetic fields below 0.4 µT and an odds ratio of around 2 was seen, indicating a twofold excess risk, for exposure above 0.4 µT [00Ahl]. The other pooled analysis included 15 studies based on less restrictive inclusion criteria and used 0.3 µT as the highest cut-point [00Gre]. A relative risk of 1.7 for exposure above 0.3 µT was reported. The two analyses are in close agreement. In contrast to these results for ELF magnetic fields, evidence that electric fields are associated with childhood leukaemia is insufficient for firm conclusions but does not suggest any risk.

In general, the animal studies, which included a number of life-time studies and studies of animals predisposed to develop cancer, and in vitro studies of cellular processes implicated in carcinogenesis, did not support the hypothesis that ELF EMF's were carcinogenic.

Subsequent to the IARC monograph a number of reports have been published concerning the risk of female breast cancer in adults associated with ELF magnetic field exposure. These studies are larger than the previous ones and less susceptible to bias, and overall are negative. With these studies, the evidence for an association between ELF exposure and the risk of breast cancer is weakened considerably and does not support an association of this kind.

In the case of adult brain cancer and leukaemia, the new studies published after the IARC monograph do not change the conclusion that the overall evidence for an association between ELF and the risk of these diseases remains inadequate.

For other diseases and all other cancers, the evidence remains inadequate.

2.3.2.4.5 Health risk assessment

2.3.2.4.5.1 Hazard identification

Biological versus adverse health effects

According to the WHO Constitution, health is a state of complete physical, mental, and social well-being and not merely the absence of disease or infirmity. Before identifying any actual health hazards, it is useful to clarify the difference between a biological effect and an adverse health effect. A biological effect is any physiological response to, in this case, exposure to ELF fields. Some biological effects may have no influence on health; some may have beneficial consequences, while others may result in pathological conditions, i.e. adverse health effects. Annoyance or discomfort caused by ELF exposure may not be pathological per se but, if substantiated, can affect the physical and mental well-being of a person and the resultant effect may be considered to be an adverse health effect.

Acute effects

ELF electric and magnetic fields can affect the nervous systems of people exposed to them, resulting in adverse health consequences such as nerve stimulation, at very high exposure levels. Exposure at lower levels induces changes in the excitability of nervous tissue in the central nervous system which may affect memory, cognition and other brain functions. These acute effects on the nervous system form the basis of international guidelines. However, they are unlikely to occur at the low exposure levels in the general environment and most working environments.

Exposure to ELF electric fields also induces a surface electric charge which can lead to perceptible, but non-hazardous effects, including microshocks.

Chronic effects

Scientific evidence suggesting that everyday, chronic, low-intensity ELF magnetic field exposure poses a possible health risk is based on epidemiological studies demonstrating a consistent pattern of an increased risk of childhood leukaemia. Uncertainties in the hazard assessment include the role of control selection bias and exposure misclassification. In addition, virtually all of the laboratory evidence and the mechanistic evidence fail to support a relationship between low-level ELF magnetic field exposure and changes in biological function or disease status. Thus, on balance, the evidence is not strong enough to be considered causal and therefore ELF magnetic fields remain classified as possibly carcinogenic.

A number of other diseases have been investigated for possible association with ELF magnetic field exposure. These include other types of cancers in children and adults, depression, suicide, reproductive dysfunction, developmental disorders, immunological modifications, neurological disease and cardiovascular disease. The scientific evidence supporting a linkage between exposure to ELF magnetic fields and any of these diseases is weaker than for childhood leukaemia and in some cases (for example, for cardiovascular disease or breast cancer) the evidence is sufficient to give confidence that magnetic fields do not cause the disease.

2.3.2.4.5.2 Exposure assessment

Electric and magnetic field exposures can be expressed in terms of instantaneous or temporally averaged values. Either of these can be calculated from source parameters or measured.

Residential exposures

In the case of residential exposure, data from various countries show that the geometric means of ELF magnetic field strengths across homes do not vary dramatically. Mean values of ELF electric fields in the home can be up to several tens of volts per meter. In the vicinity of some appliances, the instantaneous magnetic field values can be as much as a few hundreds of microtesla. Close to power lines, magnetic fields reach as much as approximately 20 µT and electric fields can be between several hundreds and several thousands of volts per meter.

The epidemiological studies on childhood leukaemia have focused on average residential ELF magnetic fields above 0.3 to 0.4 µT as a risk factor for cancer. Results from several extensive surveys showed that approximately 0.5-7% of children had time-averaged exposures in excess of 0.3 µT and 0.4-3.3 % were exposed to in excess of 0.4 µT. Calculations based on case-control studies of ELF magnetic field exposure and childhood leukaemia resulted in approximately similar ranges.

Occupational exposures

Occupational exposure is predominantly at power frequencies and their harmonics. Magnetic field exposure in the workplace can be up to approximately 10 mT and this is invariably associated with the presence of conductors carrying high currents. In the electrical supply industry, workers may be exposed to electric fields up to 30 kV m⁻¹ which induce electric fields in the body and lead to increased occurrence of contact currents and microshocks.

2.3.2.4.5.3 Exposure-response assessment

Exposure-response assessment is the process of characterizing the relationship between the exposure received by an individual and the occurrence of an effect.

Threshold levels

For some effects there may be a continuous relation with exposure, for others a threshold may exist. There will be a certain amount of imprecision in determining these thresholds. The degree of uncertainty is reflected partly in the value of a safety factor that is incorporated in order to derive the exposure limit.

Frequency-dependent thresholds have been identified for acute effects on electrically excitable tissues, particularly those in the central nervous system. These effects result from electric fields and currents that are induced in body tissues by ELF electric or magnetic field exposure. The ICNIRP [98ICN] identified a threshold current density of 100 mA m^{-2} for acute changes in functions of the central nervous system (CNS: brain and spinal cord, located in the head and trunk) and recommended basic restrictions on current density induced in these tissues of 10 mA m^{-2} for workers and 2 mA m^{-2} for members of the public. A general consideration of neural tissue physiology suggested that these restrictions should remain constant between 4 Hz and 1 kHz, rising above and below these frequencies. More recently, the IEEE [02IEE] identified threshold induced electric field strength of 53 mV m^{-1} at 20 Hz for changes in brain function in 50% of healthy adults. Effects taken into account included phosphene induction and other effects on synaptic interactions. The IEEE recommended basic restrictions on induced electric field strength in the brain of 17.7 mV m^{-1} in "controlled" environments and 5.9 mV m^{-1} for members of the public. The phosphene threshold rises above 20 Hz and therefore the basic restrictions recommended by the IEEE follow a frequency-proportional law up to 760 Hz, above which restrictions are based on peripheral nerve stimulation up to 100 kHz [02IEE]. The net effect is that the guidance recommended by the ICNIRP is more restrictive than that recommended by the IEEE at power frequencies (50/60 Hz) and above. The major factor responsible for this is the difference in cut-off frequency (20 Hz for the IEEE and 1 kHz for the ICNIRP) at which thresholds for electric field strength and induced current density begin to rise [05Rei].

No thresholds have been identified for chronic effects.

Epidemiological methods

The most common means of characterizing an exposure-response relationship in epidemiology is through the derivation of estimates of relative risk or the odds ratio per unit of exposure or across exposure categories. Most epidemiological studies have used the latter method. In summary, two pooled analyses of the studies on ELF magnetic fields and childhood leukaemia have presented dose-response analyses (see above). These analyses have been conducted both on the basis of exposure categories and of continuous exposure data. All these analyses show that the risk increase becomes detectable at around 0.3-0.4 μT . For exposure levels above these values, the data at present do not allow further analysis because of the small numbers of cases in the high exposure category.

2.3.2.4.5.4 Risk characterization and exposure guidelines

Acute effects

Exposure limits based on the acute effects on electrically excitable tissues, particularly those in the CNS, have been proposed by several international organizations. The ICNIRP (1998) guidelines for the general public at 50 Hz are 5 kV m^{-1} for electrical fields and $100 \mu\text{T}$ for magnetic fields, and at 60 Hz are 4.2 kV m^{-1} and $83 \mu\text{T}$. For workers, the corresponding levels are 10 kV m^{-1} and $500 \mu\text{T}$ for 50 Hz and 8.3 kV m^{-1} and $420 \mu\text{T}$ for 60 Hz [98ICN]. The IEEE (2002) exposure levels are 5 kV m^{-1} and $904 \mu\text{T}$ for exposure to 60 Hz EMF for the general public. For occupational groups, the IEEE levels are 20 kV m^{-1} and $2710 \mu\text{T}$ at 60 Hz [02IEE]. The differences in the guidelines, derived independently by the IEEE and the ICNIRP, result from the use of different adverse reaction thresholds, different safety factors and different transition frequencies, i.e. those frequencies at which the standard function changes slope.

Taking into account the WHO EHC document 238 [07WHO] and more recent literature reviews, ICNIRP has revised its 1998-guidelines. It is the view of ICNIRP that the currently existing scientific evidence that prolonged exposure to low frequency magnetic fields is causally related with an increased risk of childhood leukaemia is too weak to form the basis for exposure guidelines. Thus, the perception of surface electric charge, the direct stimulation of nerve and muscle tissue and the induction of retinal phosphenes are the only well established adverse effects and serve as the basis for guidance. The current

ICNIRP guidelines [10ICN2] of exposure to time varying electric and magnetic fields can be downloaded at www.icnirp.org/pubEMF.htm.

The main changes compared to the previous recommendations are:

- While in 1998 dosimetric considerations were based on simple geometrical models, the new guidelines use data from computational simulations based on anatomical detailed human body models.
- The revised basic restrictions as well as the dosimetric models used in reference levels that deviate in some areas from previous ones. There is a tendency for magnetic field reference levels to be less conservative, whereas the electric field reference levels are, with some exceptions, basically unchanged. For 50 Hz, the magnetic field reference levels are by a factor of 2 higher compared with the reference values of 1998.

Chronic effects

Scientific evidence suggesting that everyday, chronic low-intensity (above 0.3-0.4 μT) power-frequency magnetic field exposure poses a health risk is based on epidemiological studies demonstrating a consistent pattern of increased risk for childhood leukaemia. Uncertainties in the hazard assessment include the role that control selection bias and exposure misclassification might have on the observed relationship between magnetic fields and childhood leukaemia. In addition, virtually all of the laboratory evidence and the mechanistic evidence fail to support a relationship between low-level ELF magnetic fields and changes in biological function or disease status. Thus, on balance, the evidence is not strong enough to be considered causal, but sufficiently strong to remain a concern.

Although a causal relationship between magnetic field exposure and childhood leukaemia has not been established, the possible public health impact has been calculated assuming causality in order to provide a potentially useful input into policy. However, these calculations are highly dependent on the exposure distributions and other assumptions, and are therefore very imprecise. Assuming that the association is causal, the number of cases of childhood leukaemia worldwide that might be attributable to exposure can be estimated to range from 100 to 2400 cases per year. However, this represents 0.2 to 4.9% of the total annual incidence of leukaemia cases, estimated to be 49 000 worldwide in 2000. Thus, in a global context, the impact on public health, if any, would be limited and uncertain.

Concerning exposure guidelines, it is the view of ICNIRP that the currently existing scientific evidence that prolonged exposure to low frequency magnetic fields is causally related with an increased risk of childhood leukaemia is too weak to form the basis for exposure guidelines [10ICN2].

A number of other diseases have been investigated for possible association with ELF magnetic field exposure. These include cancers in children and adults, depression, suicide, reproductive dysfunction, developmental disorders, immunological modifications and neurological disease. The scientific evidence supporting a linkage between ELF magnetic fields and any of these diseases is much weaker than for childhood leukaemia and in some cases (for example, for cardiovascular disease or breast cancer) the evidence is sufficient to give confidence that magnetic fields do not cause their disease.

2.3.2.4.6 Summary

Biological effects

Exposure to power-frequency electric fields causes well-defined biological responses, ranging from perception to annoyance, through surface electric charge effects. These responses depend on the field strength, the ambient environmental conditions and individual sensitivity. The spark discharge from a person to ground is found to be painful by 7% of volunteers in a field of 5 kV m^{-1} . Thresholds for the discharge from a charged object through a grounded person depend on the size of the object and therefore require specific assessment.

High magnetic field strength, rapidly pulsed magnetic fields can stimulate peripheral or central nerve tissue, such effects can arise during magnetic resonance imaging (MRI) procedures, and are used in transcranial magnetic stimulation. Threshold induced electric field strengths for direct nerve stimulation could be as low as a few volts per meter.

The function of the retina, which is a part of the CNS, is affected by exposure to much weaker ELF magnetic fields than those that cause direct nerve stimulation. A flickering light sensation, called magnetic phosphenes, results from the interaction of the induced electric field with electrically excitable cells in the retina. Threshold induced electric field strengths in the extracellular fluid of the retina have been estimated to lie between about 10 and 100 mV m⁻¹ at 20 Hz.

The evidence for other neurobehavioral effects in volunteer studies, such as the effects on brain electrical activity, cognition, sleep, hypersensitivity and mood, is less clear. The evidence is considered inadequate.

In addition, the data do not indicate that ELF electric and/or magnetic fields affect the neuroendocrine system, neurodegenerative disorders, cardiovascular effects, the immune and haematological system, developmental and reproductive effects in a way that would have an adverse impact on human health and the evidence is thus considered inadequate.

Cancer

The IARC classification of ELF magnetic fields as “possibly carcinogenic to humans” is based upon all of the available data prior to and including 2001. New human, animal and *in vitro* studies, published since the 2002 IARC monograph, do not change the overall IARC classification of ELF magnetic fields as a possible human carcinogen. As mentioned above, the laboratory and the mechanistic evidence fail to support a relationship between low-level ELF magnetic fields and changes in biological function or disease status.

Subsequent to the IARC monograph a number of reports have been published concerning the risk of female breast cancer in adults associated with ELF magnetic field exposure. These studies are larger than the previous ones and less susceptible to bias, and overall are negative.

In the case of adult brain cancer and leukaemia, the new studies published after the IARC monograph do not change the conclusion that the overall evidence for an association between ELF magnetic fields and the risk of these diseases remains inadequate.

For other diseases and all other cancers, the evidence remains inadequate.

Risk assessment

Acute biological effects have been established for exposure to ELF electric and magnetic fields in the frequency range up to 100 kHz that may have adverse consequences on health. Therefore, exposure limits are needed. International guidelines exist that have addressed this issue. Compliance with these guidelines provides adequate protection.

Consistent epidemiological evidence suggests that chronic low-intensity ELF magnetic field exposure is associated with an increased risk of childhood leukaemia. However, a causal relationship has not been established; therefore exposure limits based upon epidemiological evidence are not recommended.

2.3.2.5 Electromagnetic fields of frequencies above 100 kHz

2.3.2.5.1 Introduction

An extensive literature has arisen during the last three decades in which a wide variety of biological effects and responses have been attributed to exposure to radiofrequency (including microwave) radiation. Many of these effects are known to result from increases in thermal loading and elevated body and tissue temperature. These particular effects and responses are well-understood and have been used by various national and international organizations as scientific bases in setting guidelines and restrictions on exposure to electromagnetic fields for human populations (e.g. [98ICN]). In addition, a variety of biological responses have been reported at very low levels of exposure which appear to be less well established. In this chapter, these two kinds of biological effects of radiofrequency (RF) radiation are discussed under two separate sections.

The first section considers the biological effects of whole body and localised heating following exposure to RF and microwave radiation at frequencies above 100 kHz. The second section describes

biological effects that are less well established, and includes cancer-related studies. The information summarised in both these sections is covered in more detail in [92Ber, 93WHO, 03AGN, 09ICN1].

2.3.2.5.2 Biological effects

The photon energy of non-ionizing electromagnetic radiation is too small to affect chemical bonding directly; at 300 GHz (the boundary between microwave and infrared radiation) the photon energy is only about 10^{-3} eV and decreases linearly with decreasing frequency. Covalent bond disruption has an activation energy 5 eV, and even hydrogen bond disruption has an activation energy of 10^{-1} eV. The electric fields induced in tissue by electromagnetic radiation result in energy absorption due to the polarisation of electrically charged structures and the flow of ions (see Sect. 2.2.2.4). It is usually assumed that the increase in linear and rotational energy is rapidly dissipated by molecular collision, resulting in generalised heating.

The heating effects of RF are well established. The total whole body heat load experienced during RF exposure is the sum of the specific energy absorption rate (SAR) and the endogenous rate of heat production. The latter varies in normal individuals from about 1 W kg^{-1} at rest, to about 10 W kg^{-1} for short periods during hard physical exercise. Power deposition within the body is never uniform; differences in the electrical properties of tissues and the reflection and refraction of radiation at the interfaces of tissues of different electrical properties can result in localised SAR “hot spots”. In addition, differences in local blood perfusion will affect heat dissipation characteristics, and some tissues are more sensitive to raised temperatures than others, and/or may be less able to affect repair.

Whole body responses

Animals, like humans, use various physiological and behavioural mechanisms in order to regulate body temperature. The responses of the thermoregulatory system to irradiation are well established and include altered rates of metabolic heat production, food intake, activity, the vasodilatation of superficial blood vessels and the behavioural selection of appropriate ambient temperatures. Thresholds for such responses have been reported in rodents and primates between about 0.3 W kg^{-1} and 5 W kg^{-1} . The magnitudes of these responses appear to depend on the frequency of irradiation used, and orientation of the exposed animal with respect to the applied electric and magnetic field. They are caused by differences in power absorption by the deeper tissues of the body, in particular the less effective stimulation of the temperature receptors in the skin results in less effective thermoregulation.

The performance of learned tasks seems particularly sensitive to RF radiation; thresholds for decreased performance in both rats and primates have been reported as lying between 2.5 and 8 W kg^{-1} ; concomitant rises in rectal temperature were around 1°C . Exposure to RF radiation can also modify the action of drugs whose effectiveness can be altered by heat-induced changes in body physiology. Changes have been recorded in the duration of barbiturate-induced anaesthesia and in the permeability of the blood-brain-barrier, but only at levels of irradiation sufficient to raise body temperature.

Other easily demonstrable effects are generally consistent with responses to non-specific stressors such as heat. The acute exposure of primates to microwaves or RF at SARs of 3 to 4 W kg^{-1} , sufficient to raise rectal temperature by 1 to 2°C , resulted in increased stress hormone (plasma cortisol) levels; similar effects have been reported in rats. In addition, microwave-induced changes have been reported in the levels of circulating white blood cells (increased levels of neutrophils and decreased lymphocyte levels) in rats and mice following thermal exposures which are similar to the changes induced by the injection of stress hormones, suggesting a common aetiology. Reported changes in white blood cell (natural killer cell and macrophage) activity have also been linked to heat induced stress.

Similar sorts of changes could be expected in humans following RF-induced increases in body temperature. It is known that most healthy people can tolerate short-term rises in body temperature by up to about 1°C , although individuals vary widely in their ability to tolerate increased body temperatures. Some individuals cannot tolerate rectal temperatures of 38°C ; others continue to perform well even at higher temperatures. However, prolonged exposure at body temperatures in excess of 38°C is known to increase the risk of heat exhaustion and reduce mental performance. Experiments have been carried out with volunteers who investigated the relationship between whole body SAR, body temperature rise and

the ensuing physiological responses. The total heat load experienced by the volunteers, resulting from the sum of the SAR and metabolic heat production, can be estimated as about 5 W kg^{-1} , which represents a typical heavy workload for many industrial jobs. Adverse environmental conditions and moderate physical exercise will reduce the tolerable level of RF or microwave energy absorption, whilst people under medication or with clinical conditions which compromise thermoregulation may be more sensitive to RF or microwave induced heating.

Localised responses

The lens of the eye is regarded as potentially sensitive to microwave irradiation because of its lack of a blood supply and consequent limited cooling ability, and its tendency to accumulate damage and cellular debris. In anaesthetized rabbits, high local temperatures induced by acute exposure of the head to microwave radiation between about 1 and 10 GHz have been shown to induce lens opacities (cataracts). The threshold SAR in the lens was between about 100 to 140 W kg^{-1} . Primate eyes were found to be less susceptible to cataract induction, possibly because they are more recessed in the skull, and so better shielded. However, thresholds for chronic exposure have not been defined. In humans, cataracts have been historically associated with chronic exposure to infrared radiation, indicating that some degree of caution should be exercised.

Testicular temperatures are normally several degrees centigrade below body temperature, and it has been known for some time that male germ cells are sensitive to elevated testicular temperatures. In humans, it has been reported that repeated heating of the testis by $3\text{--}5^\circ\text{C}$ will result in a decreased sperm count persisting for several weeks.

Heat has been shown to be teratogenic in various animal species including primates and has been associated with central nervous system and facial defects in children whose mothers developed moderate to severe hyperthermia, especially during the first trimester of pregnancy. The embryo and fetus may be particularly sensitive to RF-induced heating, since heat loss across the placenta will be less effective than heat exchange in other, well vascularised tissues. In rats, acute exposure at 11 W kg^{-1} , raising maternal temperatures to 43°C , was sufficient to induce embryo and fetal death and developmental abnormalities. Chronic exposure at 6 to 7 W kg^{-1} , usually raising maternal temperatures to between 39 and 41°C , was reported to induce growth retardation and subtle behavioural changes. In general, even prolonged exposure at less than 4 W kg^{-1} had no effect.

Surface heating

The absorption of RF and microwave radiation can be detected by temperature sensitive receptors in the skin. Power flux densities of around 300 W m^{-2} at 3 GHz have been detected experimentally during exposure for 10 seconds; radiation of higher frequencies applied for similar lengths of time have been detected at lower power flux densities because of their greater absorption by the skin. It is considered that the avoidance of the perception of skin warming for frequencies where the penetration depth is greater than the thickness of the skin (i.e. frequencies below about 10 GHz) does not provide a reliable mechanism of protection against potentially harmful exposure. The avoidance of the perception of skin warming may give adequate protection at frequencies above 10 GHz, although its general effectiveness may be reduced if other factors in the environment compete for attention.

Pulsed radiation

People with normal hearing have perceived pulse-modulated RF radiation of frequencies between about 200 MHz and 6.5 GHz; the sound has been variously described as a buzzing, clicking, hissing or popping noise, depending on modulation characteristics. Prolonged or repeated exposure may be stressful. It seems most likely that the sound results from the thermoelastic expansion of brain tissue following a small but rapid increase in temperature on the absorption of the incident energy. The perception threshold for pulses shorter than $30 \mu\text{s}$ depends on the energy density per pulse and has been estimated as about 400 mJ m^{-2} at 2.45 GHz, corresponding to an estimated specific energy absorption in the head of about 16 mJ kg^{-1} . However, a reduction in ambient noise has been reported to reduce this to about 280 mJ m^{-2} , an SA of 10 mJ kg^{-1} at 2.45 GHz.

Exposure to intense pulsed radiation has been reported to suppress the startle response and evoke body movements in conscious mice. Specific absorptions were 200 mJ kg^{-1} (for $1 \mu\text{s}$ pulses) and 200 J kg^{-1} (for

10 μ s pulses) for suppression of the startle response and evoked body movement respectively. The mechanism for these effects may be related to microwave hearing; auditory thresholds for rats are several orders of magnitude lower, about 1-2 mJ kg⁻¹ per pulse (<30 μ s) for rats.

Indirect effects

Indirect effects are those resulting from an interaction between electromagnetic fields or radiation, an external object such as a vehicle or other metallic structure, and the human body. A person touching a metallic object, such as a vehicle, situated in an RF field may perceive the discharge current and may even experience discomfort. At frequencies greater than 100 kHz, the 50th percentile threshold electric current for pain (finger contact) has been reported as about 50 mA for men, 40 mA for women and 30 mA for children. The current required to produce this effect, which takes place typically within 10 to 20 seconds, is almost independent of frequency.

Possible effects of radiofrequency radiation

The effects of exposure to RF radiation that have been described so far are well established and fairly well understood. There are in addition a large number of biological effects that have been reported in cell cultures and in animals, often in response to relatively low field levels, which are not well established but which may have health implications and are the subject of much ongoing research. These include research on the effects of RF radiation on carcinogenic processes, on the existence of specific frequency and amplitude "window" effects, and on the effects of low level pulsed RF radiation.

Much biological research has focused on carcinogenic processes. Most experimental evidence suggests that RF radiation is not mutagenic and so is unlikely to initiate tumours. Recent studies have mostly concentrated on aspects of cellular metabolism relevant to tumour promotion or on an enhancement of the incidence of spontaneous or chemically-induced tumours in animals. However, there is little persuasive evidence to suggest that RF radiation is able to influence any of the accepted stages in carcinogenesis: few clearly reproducible effects are apparent. The results of the large scale animal studies are mostly equivocal, although tumour progression may be enhanced at thermally significant levels. Some data challenge the conventional assumption that the magnitude of an effect increases with increasing exposure. In general, however, the effects following exposure to low level radiation have not been well established and are not considered able to provide a basis for restrictions on human exposure.

2.3.2.5.3 Risk assessment and exposure guidelines

2.3.2.5.3.1 Introduction

The biological effects of exposure to RF electromagnetic fields have been the subject of investigations for more than 50 years. It is well known that at sufficiently high intensities radiofrequency energy can produce high absorption rates in tissue and engender adverse thermal effects that affect functioning in the human body. The causal relationship between RF exposure and biological response begins with RF exposure of the biological system. The incident or applied RF electric and magnetic fields must be coupled into cells and tissues, and energy must be deposited or absorbed in the biological system in order for the system to respond in some manner, regardless of any mechanism accountable for a response. Some effects may result from the induced fields, and others may be associated with elevation of tissue temperature from RF energy deposited or absorbed in biological systems. Thus, in order to achieve any biological response, the electric, magnetic, or electromagnetic field that is exerting its influence must be quantified and correlated with the observed phenomenon.

A complicating factor of the interaction is that the same exposure or incident field does not necessarily provide the same field inside biological systems of different size, species, or constitution. Thus, unrestricted exposure of the body to a plane wave at a given strength can have outcomes far different from those of partial-body or localized exposure at the same field strength. The spatially averaged field strength, depending on the region of space over which the fields are averaged, may vary

widely for a given body. Therefore, an important task in assessing the health and safety of RF exposure is the determination of induced fields in biological tissues.

2.3.2.5.3.2 Quantifying induced field and SAR

The induced electric fields in biological tissue are functions of body geometry, tissue permittivity and conductivity, and source configuration and frequency. There exists a variety of experimental and computational methods for quantifying induced fields and SARs inside biological bodies. Some are appropriate for homogeneous phantoms and others are more suitable for heterogeneous anatomic models.

At lower frequencies, e.g., 100 kHz or 10 MHz, where the wavelength of RF radiation is at least an order of magnitude longer than the dimensions of the human body, field behaviour inside the body is characterized by near-zone reactive field and is quasi-static in character. The electric and magnetic fields become decoupled and act separately and additively inside tissue medium [88Ber]. For all practical purposes, the induced fields can be obtained by combining the two independent electrostatic and magnetostatic solutions of the electromagnetic field theory, for example, an externally applied uniform electric field gives rise to a uniform induced electric field in the same direction, inside a body without abrupt changes in cross sections, but reduced in strength by a factor inversely proportional to the dielectric constant and is independent of body size. The magnetically induced electric field amplitude inside the body is approximately given by

$$E = \omega B r / 2 = \pi f r \mu H, \quad (2.3.2.1)$$

where $f = \omega / 2\pi$ is the frequency, μ is permeability, r is the equivalent, body radius, B is the magnetic flux density, and H is the strength of the magnetic field component. A uniform magnetic field produces an internal electric field that increases in proportion with the distance away from the body centre. Thus, magnetically induced electric field, i.e., inductive coupling, would dominate inside a biological body except for tissue bodies that are 1 mm or less in size. A similar scenario exists in the near-zone reactive region of all antennas and radiating systems.

Numerical methods have matured to a level that they are being increasingly used for dosimetric calculations. Applications to date have modelled the source as simple dipoles, practical antennas for cellular mobile telephones or for biomedical applications, or plane waves far from the antenna and have used both simple geometric and detailed anatomic representations of the head, neck, thorax, and the entire body. Some of these models, especially the anatomical ones, can take into account the internal organs, tissue heterogeneities, and complex dielectric permittivity, which can influence induced field and SAR.

Several computed tomography (CT) or magnetic resonance imaging (MRI) based models of the human adult and animal bodies have been used for SAR and field calculations. The computerized models have a spatial resolution of 1-3 mm. Questions have been raised concerning the comparability of computational results using different models. A particularly vexing problem is the influence of tissue heterogeneity and the volume element size on computed SAR distribution in these models. These concerns suggest the need for standardization in various modelling efforts to simulate realistic exposure in order to minimize the difficulty encountered in dosimetric determinations and comparisons. It is noteworthy that available info-action from literature suggests that the integrated or whole-body SAR is similar for a homogeneous or heterogeneous model. The maximum SAR and its distribution are closely related to the distance of the radiating element from the body surface and the current distribution on the antenna in personal communication devices [07Lin].

2.3.2.5.3.3 Basic restrictions and reference levels

Guidelines for safe exposure to RF radiation have been developed for about half a century. At present the ICNIRP guidelines [98ICN] for limiting human exposure to RF radiation have been adopted by more than 50 countries worldwide. These guidelines are relied on the interpretation of published scientific papers selected from the peer-reviewed literature that were deemed to have biological, engineering, and scientific

validity. The limits established were intended to prevent excessive localized tissue heating, such as to the eyes and testes. Considerations also were given to exposures that could lead to disruption of work in trained rodents and primates, and other adverse biological effects resulting from either partial-body or whole-body exposures that could produce temperature rises on the order of 1°C in humans and laboratory animals [03Lin].

The published scientific literature will be reviewed periodically by several national and international expert bodies in order to check whether the scientific base for the guidelines has changed. A recent report of the Scientific Committee on Emerging and Newly Identified Health Risks (SCENIHR) of the European Commission, based on more than 200 scientific papers, published the following statement: “Based on current evidence the main conclusion remain that radio frequency fields used in wireless communication technologies are unlikely to lead to an increase in cancer in the human population at large. However, further studies are needed to clarify if long-term exposure to mobile phones (well beyond 10 years) increases cancer risk for an individual using a mobile phone frequently and to examine the effects on children” [09SCE].

In addition, in a recent statement ICNIRP emphasised its opinion that the scientific literature published since the 1998 guidelines has provided no evidence of any adverse effects below the basic restrictions and does not necessitate an immediate revision of its guidance on limiting exposure to high frequency electromagnetic fields, and therefore confirmed the 1998 basic restrictions in the frequency range 100 kHz-300 GHz until further notice [09ICN3].

Basic restrictions

Basic restrictions are established on the basis of recognized health effects and biological responses, after incorporation of appropriate safety factors, to limit human exposure. Safety factors have been incorporated in order to take into account scientific uncertainty and also biological variability in the human population. In general, for RF frequencies below 10 MHz, the basic restrictions are expressed in terms of induced current density for preventing effects in excitable tissues such as nerves and muscles. At intermediate frequencies (100 kHz to 100 MHz), induced current density and SAR are selected for prevention of local tissue heating and whole-body heat stress. At higher frequencies (100 MHz to 10 GHz), SAR is used for prevention of whole-body heat stress and local tissue heating, and at very high frequencies (10 to 300 GHz), the basic restrictions are expressed in terms of power density for prevention of excessive tissue heating near or at the body surface, especially for acute exposures.

Reference levels

Since basic restrictions are often specified in quantities that are impractical to be measured in common exposure environments, other quantities are introduced for practical exposure assessment purposes in order to determine whether the basic restrictions are likely to be exceeded. These quantities are termed reference levels [98ICN]. They correspond to basic restrictions under worst case exposure conditions as determined by one or more of the following physical quantities: incident electric field strength, incident magnetic field strength, magnetic flux density, power density, limb current, contact current, and, for pulsed fields, peak electric field, peak power density, and specific energy absorption.

Pulsed Fields

The interaction of pulsed fields with biological systems can be very different from exposure by continuous waves. Reference levels have been adopted for pulsed fields in the international guidelines to prevent such adverse biological effects in humans as dielectric breakdown in tissues and induced unconsciousness, as well as annoyance invoked by the microwave auditory phenomenon.

The current ICNIRP guidelines of exposure to electromagnetic fields can be downloaded at www.icnirp.org/pubEMF.htm.

2.3.2.5.4 Summary

The mechanisms by which RF exposure heats biological tissue are well understood and the most marked and consistent effect of RF exposure is that of heating, resulting in a number of heat-related physiological and pathological responses in human subjects and laboratory animals. Heating also remains a potential confounder in *in vitro* studies and may account for some of the positive effects reported.

Children are known to thermoregulate as well as adults in response to exercise and/or hot environments, but may be more vulnerable to dehydration. Similar cardiovascular responses to RF-induced heating such as increased skin blood flow occur in laboratory animals. However, animals are less effective at dissipating excess heat than humans, being in general less able to increase skin blood flow and sweat although heat loss can also occur via other mechanisms such as panting. The evidence of volunteer studies suggest that cognitive functions can be adversely affected by whole-body heat stress, resulting in increased levels of unsafe behaviour and reduced task performance. However, laboratory animals show a consistent reduction in the performance of learned behaviours when RF-exposure increases core body temperatures by about 1°C or more. Similar RF-induced rises in body temperature also result in significantly enhanced plasma corticosterone or cortisol levels in rodents or primates respectively and transient changes in immune function and haematology, generally consistent with the acute responses to non-specific stressors. However, these thermal effects have not been systematically explored in RF volunteer studies.

Most recent studies of human subjects, including adults, children and adolescents, have focused on the possible effects of essentially non-thermal exposures to mobile phone type RF, often simulating mobile phone use only involving localized exposure of part of the head. A number of non-thermal interaction mechanisms have been proposed but to date none have been experimentally verified.

Concerning cancer-related effects, the recent *in vitro* and animal genotoxicity and carcinogenicity studies are rather consistent and indicate that such effects are unlikely at SAR levels up to 4 W kg⁻¹. With regard to *in vitro* studies of RF effects on non-genotoxic end-points such as cell signalling and gene/protein expression, the results are more equivocal, but the magnitudes of the reported RF radiation induced changes are very small and of limited functional consequences. The results of studies on cell proliferation and differentiation, apoptosis and cell transformation are mostly negative.

It is clear that, like humans, animals can hear the pulsed RF characteristic of radar above given thresholds through a thermoelastic expansion mechanism. Studies of the effects of high peak power RF pulses and ultrawide band (UWB) RF has been somewhat diverse and sporadic.

Thermally significant RF exposure can impair male fertility and cause increased embryo and fetal losses and increase the incidence of fetal malformations and anomalies. Such effects have not been consistently shown at exposure levels that do not induce temperature elevation of 1°C or more.

The studies that have addressed postnatal developmental indices or behaviour after prenatal exposure to low level RF radiation have generally reproduced lack of effects. Effects resulting from long-term exposure during the development of juvenile animals have been addressed in only a few studies, and the data are insufficient for conclusions.

Cataract in the eyes of anesthetized rabbits remains a well-established thermal effect of RF exposure. However, primates appear less susceptible to cataract induction than rabbits, and opacities have not been observed in primates following either acute or prolonged exposures.

Guidelines for safe exposure to RF radiation have been developed for about half a century. At present the ICNIRP guidelines for limiting human exposure to RF radiation have been adopted by more than 50 countries worldwide. The limits established were intended to prevent excessive localized tissue heating, such as to the eyes and testes. Considerations also were given to exposures that could lead to disruption of work in trained rodents and primates, and other adverse biological effects resulting from either partial-body or whole-body exposures that could produce temperature rises on the order of 1°C in humans and laboratory animals.

In the established exposure guidelines, basic restrictions are established on the basis of recognized health effects and biological responses, after incorporation of appropriate safety factors, in terms of biologically effective dosimetric quantities like SAR and current density. SAR and current density and their determinations are tissue-type dependent and require a region of specific tissue mass for averaging and for correlation with any induced biological response.

Reference levels expressed in terms of physical quantities are introduced for practical exposure assessment purposes. They correspond to basic restrictions under worst case exposure conditions as determined by one or more of the following physical quantities: incident electric field strength, incident magnetic field strength, magnetic flux density, power density, limb current, contact current, and, for pulsed fields, peak electric field, peak power density, and specific energy absorption.

2.3.2.6 Optical radiation and lasers

2.3.2.6.1 Introduction

Because of its limited penetration in biological tissues the direct hazardous effects of optical radiation are limited to the skin and to the eyes. Humans have been required to adapt to an optical radiation environment dominated by solar radiation. The eyes are normally protected from acute retinal injury, which can be caused by staring at the sun, by involuntary aversion responses. These responses protect the retina from visible radiation and from IR-A (770-1400 nm) radiation both of which are focused onto the retina forming a resolved image of the sun. The cornea is normally physically protected from the acute effects of solar UVR by the recessed protective position of the eye in its socket and by the glancing incidence of the radiation. Exposed skin is at risk from both the acute and chronic effects of solar radiation. Even after adaptation, solar radiation may still be hazardous and it undoubtedly causes more adverse health effects than any human made source of optical radiation. However, with scientific and industrial advances human-made sources of optical radiation have been developed with similar or even greater potential for injury.

The major physical interactions of optical radiation with biological tissues include specular and diffuse reflection, refraction, scattering and absorption. Only energy absorption can result in the physical and chemical (photochemical) changes that result in a biological effect.

In the following sections the biological effects of exposure to optical radiation are described separately for ultraviolet radiation, visible and infrared radiation and laser radiation. Since the ultraviolet range is the most dangerous part of non-ionizing radiation, this part is explained in more detail. For further reading the following reviews are mentioned: [80Sli, 82WHO, 94WHO, 02AGN, 04ICN, 07ICN].

2.3.2.6.2 Biological effects of ultraviolet radiation

2.3.2.6.2.1 Biologically significant exposure dosimetry

In photobiology, the concept of a biologically effective dose is of critical importance. Since not all wavelengths of UVR are equally effective in producing a biological effect, an action spectrum $A(\lambda)$, which defines the relative effectiveness of different wavelengths, is determined. This relative response curve is generally normalized to provide a maximal value of 1.0 at the wavelength of maximal tissue sensitivity. When considering health effects of UVR, an effective exposure rate (i.e., irradiance) E_{eff} (or the exposure summed over time, i.e., the effective radiant exposure H_{eff}) is calculated by spectral weighting as follows: the spectral irradiance E_λ at the surface of the exposed biological tissue is mathematically weighted against the action spectrum of the biological response which is a function of the wavelength, $f(\lambda)$, across the relevant spectrum (e.g., from 200 nm to 400 nm) as is shown above in Sect. 2.2.2.5

2.3.2.6.2.2 Biological effects on the skin

Ultraviolet radiation is absorbed to varying degrees by all constituents of living organisms and so, in the epidermis by nucleic acids (DNA, RNA), proteins, and chromophores dispersed in the cytosol and membranes. Interactions with biomolecules will result in absorption of specific UV wavelengths by

corresponding molecular structures and result in production of excited state. The primary product generated by UV absorption is a reactive species in an excited state or free radical.

The peak absorption of DNA occurs at around 260 nm with a sharp drop in absorption through the UVB range (several orders of magnitude). No absorption is detected for wavelengths longer than 325 nm.

DNA strand breaks are induced by UVB, UVA and shortwave visible radiation range. The phenomenon is dependent on oxygen. Also absorption of UVA may result in DNA-protein cross-links. All DNA lesions should be repaired before the cell is engaged in division. Several mechanisms are involved and gene inactivation may result from mutation in its structure.

2.3.2.6.2.3 Acute effects on the skin

Sunburn

“Sunburn” is an acute injury following excessive exposure to UVR and is most pronounced for lightly pigmented skin types. Sunburn is actually not caused by heat or caustic chemicals, but is the result of a phototoxic (actinic) effect in the skin. Unlike the other burns, sunburn is not immediate. Skin redness reaches a maximum at about 8-12 hours after exposure and fades within a few days. The red appearance of the skin (erythema) results from an increased blood content near the skin’s surface. The non-adapted (“untanned”) skin of very lightly pigmented Caucasian subjects will normally show signs of a mild reddening after about 4 hours following only a half-hour exposure to midday summer sunshine in mid-latitudes. Higher doses may result in pain and skin swelling (edema) with blistering, and after a few days, peeling. Sunburn sensitivity varies substantially with skin complexion and colour, and this is reflected in the solar exposure time required to induce a sunburn reaction – from 15-30 minutes of midday summer sunshine to 1-2 hours exposure for moderately pigmented skin; and those with darkly pigmented skin may not clearly show sunburn for a full day exposure. Skin specialists frequently group individuals into one of six sun-reactive skin types ([Table 2.3.2.2](#)), and these skin types fall into three more significant groups based upon how well individuals produce the pigment, melanin, in their skin (see next section) [[99ISO, 94Dif, 82Par](#)].

Tanning and adaptation of the skin

The wide range of susceptibility to solar exposure among phototypes ([Table 2.3.2.2](#)) is due largely to the two types of melanin (eumelanin and phaeomelanin) [[04You1](#)]. Phaeomelanin absorb UV photons and produce reactive oxygen species (ROS) which are phototoxic. Eumelanin (stable free radical) absorb UV photons and scavenge free radicals and are photoprotective. All individuals produce eumelanin and phaeomelanin in different ratios according to genetic makeup and as a consequence present large differences in solar sensitivity and skin cancer incidence. In addition, darker skin types have more efficient DNA repair than the skin phototypes I and II [[02She](#)].

Skin adaptation from frequent UVR exposure is not only characterized by the obvious effect of skin darkening (“tanning” or “melanogenesis”), but also by skin thickening. Thickening of the outermost layers of the skin (epidermis and stratum corneum) takes place as an adaptation to UVB-related damage. This can be a 3 to 5-fold thickening of the stratum corneum within one to seven weeks after several exposures to UVB, and returns to normal about one to two months after ceasing exposure. This thickening after sun exposure leads to a significant increase in UV protection by a factor of five or greater, and in lightly pigmented skin types, thickening is probably more important than tanning in providing protection. The thickening of the skin after prolonged tanning protects sensitive cells (basal keratinocytes, melanocytes) by absorbing UVB radiation before they reach the basal layer of the epidermis. After some shedding (peeling) of the stratum corneum, the basal layer can be directly stimulated by UVB and thus the thickening or protective processes recur and reach a steady state. However, in darkly pigmented individuals it is likely that skin pigmentation is the most important means of protection against UVR.

The wavelengths of the radiation that induce tanning are very similar to those of radiation producing erythema. [Table 2.3.2.2](#) describes the range of skin types and sensitivity to UVR effects. Subjects with sun-reactive, melano-compromised (skin types I and II) are poor tanners compared to those with melano-

competent (skin types III and IV) who tan well. Melanogenesis can be stimulated in individuals who tan well with solar UV doses that were considerably below the erythema doses in the UVA region.

Some chemicals can enhance the sensitivity of the skin to UVR (most notably, UVA) through what is known as “phototoxicity”. Photosensitizers like certain drugs, plant materials, perfumes and cosmetic constituents, dyestuffs, polycyclic hydrocarbons in wood preservatives, coal tars, pitch and pollutants, sunscreen and printing ink materials can enter the skin from the surface or through other routes to produce phototoxic reactions. Photosensitizers can be found in domestic work environments, outdoor workplaces, and in industrial working places. In addition, the strongest photosensitizers are often administered for medical purposes, and workers exposed to UVR should be aware of this potential.

Table 2.3.2.2. Classification of skin types based on their susceptibility to sunburn in sunlight and their ability to tan. Adapted from [07ICN].

Skin	Sun sensitivity	Sunburn susceptibility	Tanning achieved	Classes of phototype individuals
I	Very sensitive	Always sunburn: < 2 SED	No tan	Melano-compromised
II	Moderately sensitive	High: 2-3 SED	Light tan	Melano-compromised
III	Moderately insensitive	Moderate: 3-5 SED	Medium tan	Melano-competent
IV	Insensitive	Low: 5-7 SED	Dark tan	Melano-competent
V	Insensitive	Very Low: 7-10 SED	Natural brown skin	Melano-protected
VI	Insensitive	Extremely low : > 10 SED	Natural black skin	Melano-protected

2.3.2.6.2.4 Chronic effects on the skin

Photoaging

Photoaging from occupational exposure has traditionally been particularly observed in fishermen and farmers in sun exposed sites such as the face and the back of the neck and hands. The clinical signs of a photo-aged skin are dryness, deep wrinkles, accentuated skin furrows, sagging, loss of elasticity, mottled pigmentation and the development of tiny but highly visible, superficial blood vessels (telangiectasia). These characteristics reflect profound structural changes in the dermis. It is not yet clear which wavelengths are most responsible for photoaging, but some research studies point to solar UVA and even infrared radiation exposures as contributing factors.

Skin cancers

The three common forms of skin cancer, listed in ascending order of severity are: basal cell carcinoma (BCC), squamous cell carcinoma (SCC) and malignant melanoma (MM). Around 90% of skin cancer cases are of the non-melanoma variety (BCC and SCC) with BCCs being approximately four to eight times (depending on latitude) as common as SCJs. Exposure to UVR is considered to be a major etiological factor for all three forms of cancer [92IAR]. For basal cell carcinoma and malignant melanoma, neither the wavelengths involved nor the exposure pattern that results in risk have been established with certainty, whereas for SCC, both UVB and UVA are implicated and the major risk factors seem to be cumulative lifetime exposure to UVR and a poor tanning response. The risk of developing skin cancer varies greatly with skin type, and more than 90% of skin cancers are found in melano-compromised persons (Table 2.3.2.2). These individuals should be advised to choose occupations where UVR exposure is minimal. When UVR exposure is unavoidable, protective measures (hats, clothing) is strongly recommended. Therefore, persons who readily sunburn are also more prone to develop skin cancer. Precursor lesions of SCC known as actinic keratoses are common in fair-skinned outdoor workers by the age of 50 to 60 (depending upon latitude).

2.3.2.6.2.5 Systemic effects

Production of Vitamin D₃

The best-established beneficial effect of solar ultraviolet radiation on the skin is the synthesis of vitamin D₃ [89Web]. Solar radiation in the UVB waveband photochemically converts 7-dehydrocholesterol in the epidermis to previtamin D₃, which is then converted to vitamin D₃. Sunlight regulates and limits further production of vitamin D₃ in the skin to preclude a toxic level. During midday hours in the tropics or during the spring and summer in more temperate climates, only brief sub-erythema exposures to sunlight are required to synthesize vitamin D₃. In this case, less than 15 minutes of solar UVR exposure on the hands, arms and face fulfills this requirement.

Vitamin D is known to be essential for the body's proper uptake of calcium, which is important for bone and musculoskeletal health. There is also evidence of an increased risk of autoimmune diseases, including multiple sclerosis, type-I diabetes and rheumatoid arthritis with low vitamin D intake or low UV exposure, respectively. Some recent epidemiological studies suggest a link between a number of cancers and low vitamin D levels. At present, however, the evidence is insufficient to establish a causal relationship and the questions on what is the optimal level of vitamin D and what is the amount of UV needed to maintain an adequate vitamin D level still remain difficult to answer (reviewed in [06ICN]).

UVR immune effects

Through photochemical reactions, UV radiation can alter various organic molecules in the skin and these molecules may become dysfunctional and "foreign" to the skin. An important task of the body's immune system is to seek out and destroy any "foreign intruders", and it is therefore conceivable that UV radiation could trigger unwanted immune reactions against the skin (this might actually cause "sun allergy") [92Coo]. It has, however, been established that UVB exposure reduces specific immune reactions in the skin of healthy people and this is likely to constitute a healthy response to avoid any undesired immune reaction against the UVR exposed skin.

2.3.2.6.2.6 Effects on the eye

Exposure of the eye to UVR is associated with a variety of disorders, including damage to the eyelids, cornea, lens and retina.

Radiation with wavelengths shorter than 290 nm is almost entirely absorbed by the cornea. Further, radiation in the range 300-370 nm is almost entirely attenuated in the lens. There is a strong increase of UVR attenuation by the lens with increasing age. If the lens is removed (cataract surgery) without implantation of an UVR absorbing lens or if there is no lens (i.e. aphakia, which is rare), a significant fraction of the incident UVR (290-400 nm) may reach the retina.

The cornea does not adapt to repeated exposures like the skin (by thickening of the stratum corneum and epidermis). Therefore, the cornea is equally vulnerable day after day to the same exposure to UVR.

The effects of UVR on the eyelids are equivalent to those described for the skin. For this reason only adverse effects of UVR to the eye will be considered here.

Since the transparent media of the eye do not have any melanin pigment (as in the skin), there is no correlation between the UVR sensitivity of the eye and skin type.

Acute effects

An unprotected eye exposed to UVR from the sunlight reflected from light sand or snow during one day may accumulate a sufficient dose to cause an adverse effect in the cornea of the eye. As with sunburn of the skin, the symptoms are delayed for several hours. Within six hours such an exposure gives rise to a gradual transition symptoms from a feeling of itchiness, "sand in the eye" sensation, increased tearing, to severe pain and photophobia (light sensitivity). This is caused by an inflammatory reaction in the cornea and conjunctiva known as photokeratoconjunctivitis, which leads to a swelling and loss of the superficial cells in the cornea and the conjunctiva. Within 24-48 hrs, the pain decreases and the light sensitivity disappear. This condition is popularly referred to as "snowblindness" or "welders flash".

In addition to corneal injury, laboratory studies have demonstrated acute cataract formation from UVR at wavelengths greater than 310 nm emitted by artificial or laser sources [02Hoc].

Chronic effects

Development of cataract, a clouding of the lens that disturbs vision, is part of the natural ageing process. Epidemiological data show an increased risk for cortical cataract with UVB exposure from the sun [02Sas1]. The prevalence of the blinding disease of cataract world-wide exceeds 50 million [94WHO]. Animal experiments have clearly shown that UVR exposures produce cataracts, but experts disagree on the degree of importance played by environmental solar exposure [02Sl].

2.3.2.6.3 Biological effects of visible and infrared radiation

Skin injury

Photosensitized injury of the skin by visible radiation is possible as a result of the presence of both endogenous and exogenous photosensitizers such as serum bilirubin and phenothiazine. Although this effect is far less likely to be caused by light than by UV radiation, it may occur with the ingestion of certain photosensitizing molecules in food or medicines [82Dif1]. For example, the action spectrum for porphyria frequently has secondary peaks at about 400 nm and 500 nm.

Thermal injury of the skin is rare from most non-laser sources and is highly dependent upon the source size and the initial skin temperature (usually 22–25°C compared with 37°C for the retina). Very high irradiances are needed to produce thermal injury within the pain reaction time (< 1s). In typical industrial situations, whole-body heat stress tends to limit the duration of exposure to optical radiation, keeping it below the threshold for thermal damage to the skin. Hence only pulsed, or very brief, exposures to very high irradiances pose a thermal hazard to the skin.

Effects on the eye

The principal retinal hazard from viewing bright light sources is photoretinitis, e.g., solar retinitis with an accompanying scotoma from staring at the sun. Solar retinitis was once referred to as “eclipse blindness” and associated “retinal burn”. It is now clear that photoretinitis is the result of a photochemical reaction following exposure of the retina to shorter wavelengths in the visible spectrum, i.e., violet and blue light, and it is now frequently referred to as blue-light retinal injury [80Sl]. Only the very high irradiances of sources such as a xenon-arc flash lamp, a nuclear flash or a laser are capable of producing thermal injury of the retina.

Laboratory studies suggest that photochemical injury from chronic low-level exposure is related to absorption by the retinal pigmented epithelium and choroids of short wavelength light in the 380–520 nm region [78Ham]. In deriving the exposure limits for photoretinitis, it was decided that the acute photoretinitis action spectrum of Ham (1989) was the most appropriate for determining hazards from blue light [89Ham, 97ICN]. Plausible exposure durations for viewing intense light sources are similar to the duration of the aversion response (<0.20 s). The potential hazard of longer durations is essentially nullified by involuntary eye movements (which distribute the light energy over a much greater area of the retina) and by behavioural reactions such as movement of the head.

The mechanisms involved in acute retinal injury as a function of retinal image size (i.e., both for minimal-spot-size, intrabeam viewing and for extended sources) in the wavelength region 380–1400 nm are well understood. Although there are no clear boundaries between injury mechanisms, certain mechanisms dominate depending on the spectral region and the exposure duration. For short-duration exposures (less than a few seconds), the damage is due to thermal injury at threshold. The mechanical disruption of tissue caused by ultra-short laser pulses does not occur with current non-laser sources and is therefore not considered here. Photochemical, rather than thermal, effects dominate only in the wavelength region below approximately 600–700 nm for exposure times in excess of 10 s. At infrared wavelengths where photochemical effects have not been detected, thermal effects still dominate for exposure times longer than 10 s. Radial heat flow produces a strong dependence of retinal injury threshold on retinal image size. Eye movements (at exposure times greater than about 0.1 to 10 s) also create spot-size dependence for photochemical retinal injury [89Sl].

Different ocular structures are affected by different infrared spectral bands: for wavelengths up to 1350-1400 nm, the ocular media transmit energy to the retina. At longer wavelengths, the anterior segment of the eye absorbs incident energy. The infrared radiation that is absorbed by the anterior segment (the cornea, aqueous, and lens) can produce clouding of the cornea and lens when the corresponding thresholds are exceeded. Exposure limits are set to protect both against acute as well as chronic exposure.

For IR-B or IR-C exposure of the eye, Okuno (1994) showed that the temperature rise is the largest at the corneal surface and decreases gradually towards the retina [94Oku]. He also showed that the temperature rise is the quickest at the corneal surface and is slightly delayed at deeper ocular structures. Focal radiation in the lens can occur with the use of certain ophthalmic-instrument light sources [05Oku]. The cornea is extremely sensitive to thermal stimulus and this will tend to limit hazardous infrared exposure.

2.3.2.6.4 Biological effects of laser radiation

Effects of UV radiation

Short-wavelength UV-B and UV-C radiation is absorbed within the cornea and conjunctiva, whereas UV-A radiation is absorbed largely in the lens [82WHO]. Exposure to “actinic” UV (UV-B and UV-C) laser radiation may produce the acute effects of erythema (reddening of the skin), photokeratitis (corneal inflammation), and conjunctivitis. Typically, 1000-fold greater intensities or durations of UV-A exposure are required to produce photokeratitis and erythema by a photochemical mechanism. Thermal injury to the skin or the lens and cornea from UV-A exposure has been demonstrated for short pulse durations but has not been demonstrated experimentally for UV-A exposure durations greater than 1 ms. With longer exposures, photochemical effects dominate. For photokeratitis, peak sensitivity is believed to be around 270 nm, with a decrease in the action spectrum in each direction. The peak of the erythema action spectrum varies from 200 to 300 nm depending upon the definition of the degree of severity and the time of assessment of the effect. In the actinic-UV region, the cornea is not substantially more sensitive to injury than untanned lightly pigmented skin, but corneal damage is much more disabling (and painful). Repeated exposure of the skin results in tanning and thickening of the stratum corneum, which provides increased natural protection, the same is not true of the cornea. Although there is greater absorption of UV-A than of UV-B in the lens, it now seems likely that cataract formation is primarily due to excessive UV-B exposure.

Effects of visible and IR-A radiation

The primary effect on the eye of visible and IR-A radiation (400-1,400 nm) is damage to the retina. For a point source of light or for intrabeam viewing, the optical gain in irradiance from the cornea to retina is approximately 100000. Most of the radiation that reaches the retina is absorbed by the pigmented epithelium and the underlying choroids (which supplies blood to much of the retina). The photopigments in the retina absorb only a small fraction of the incident radiation. Injury to the skin in this spectral region results from temperature rises exceeding 45°C; photosensitization of the skin to visible light is rare.

Data are available for the radiation thresholds at which biological effects become manifest, and the mechanisms of retinal injury as a function of retinal image size (i.e., both for minimal-spot-size, intrabeam viewing and for extended sources) in the wavelength region 400-1400 nm are understood. Although there are no definite boundaries between injury mechanisms, certain mechanisms dominate according to the exposure duration at threshold. For short-duration exposures (less than a few seconds) the damage is due to thermal injury. Q-switched pulses lasting of the order of 10 ns will also cause mechanical disruption at levels somewhat higher than the threshold for simple thermal coagulation of tissue. Photochemical, rather than thermal, effects predominate only in the wavelength region from 400 nm to approximately 550-700 nm for lengthy exposure times (more than 10 s). At IR-A wavelengths where the photochemical effect apparently disappears, thermal effects still dominate for exposure times in excess of 10 s. Radial heat flow produces a strong dependence of retinal injury threshold on retinal image size [89Cou]. At short wavelengths in the visible spectrum, eye movements become an important factor

for exposure times greater than about 0.1-10 s, and photochemical injury of the retina also becomes dependent on spot size [89Sli].

Effects of infrared (IR-B and IR-C) radiation

In the IR-B and IR-C regions of the spectrum (wavelengths greater than 1.4 μm), the ocular media are opaque because of absorption of the radiation by the water component. Thus, in these infrared regions, radiation causes damage primarily to the cornea, although lens damage has also been attributed to wavelengths below 3 μm (IR-A and IR-B). The IR damage mechanism appears to be thermal, at least for exposure durations greater than 1 μs ; for pulses of shorter duration the mechanism may be thermomechanical. The CO₂ laser (10.6 μm), the Nd:YAG laser (1.06 μm), and the holmium laser (12.1 μm) that are used in surgical applications are typical of IR radiation sources that cause thermal injury to tissue. In the IR-C region, as in the UV, the exposure threshold for damage to the skin is comparable with that for damage to the cornea. However, damage to the cornea is likely to be of greater concern because of the adverse impact on vision.

2.3.2.6.5 Derivation of exposure limits

The exposure limits for both conventional light sources and lasers (pulsed and continuous) are based on human and animal thresholds for ocular injury. According to current knowledge, and because the expression of fundamental limits for use with lasers may be simplified, the exposure limits and safety factors for the two sources necessarily differ. The use of all laser limits as if they constituted an action spectrum to spectrally weight a broad-band source would result in a very conservative hazard evaluation, since multiple effects from different wavelength bands would be summed together. Because the ocular exposure limits were derived to protect different structures (e.g., cornea, lens, retina), the different biological effects are not additive.

The biological effects induced by incoherent and coherent optical radiation are believed to be similar for any given exposure site, area, and duration of exposure in the same spectral region, but thresholds of injury can differ for very narrow spectral bandwidths. Therefore, laser radiation must be treated as a special case because of the spectral purity and radiant intensities achieved only by lasers. Data for many biological effects are based on broad-band optical sources and are therefore not directly applicable to the highly monochromatic emissions of lasers. The degree of quantitative uncertainty in relating a biological threshold derived from broad-band and narrow-band sources to laser exposure has frequently necessitated the use of additional safety factors in deriving the exposure limits for lasers that are unnecessary for broad-band optical sources.

In developing exposure limits for broad-band optical sources, action spectra were specified and used to spectrally weight source emissions to derive a “biologically effective radiance or irradiance”. This provides the most accurate hazard assessment; exposure limits can then be specified in terms of exposure duration and other relevant parameters so that all sources are evaluated with the same risk criteria.

ICNIRP guidelines of exposure to

- Ultraviolet Radiation of Wavelengths between 180nm and 400 nm (Incoherent Optical Radiation)
- Broad-band Incoherent Optical Radiation (0.38 to 3 μm)
- Laser Radiation of Wavelengths between 180 nm and 1000 μm

can be downloaded at www.icnirp.org/pubOptical.htm. The most recent ICNIRP publication is a statement on protection of workers against ultraviolet radiation [10ICN1].

2.3.2.6.6 Summary

Because of its limited penetration in biological tissues the direct hazardous effects of optical radiation are limited to the skin and to the eyes. Humans have been required to adapt to an optical radiation environment dominated by solar radiation. The eyes are normally protected from acute retinal injury,

which can be caused by staring at the sun, by involuntary aversion responses. These responses protect the retina from visible radiation and from IR-A (770-1400 nm) radiation both of which are focused onto the retina forming a resolved image of the sun. The cornea is normally physically protected from the acute effects of solar UVR by the recessed protective position of the eye in its socket and by the glancing incidence of the radiation. Exposed skin is at risk from both the acute and chronic effects of solar radiation. Even after adaptation, solar radiation may still be hazardous and it undoubtedly causes more adverse health effects than any human made source of optical radiation. However, with scientific and industrial advances human-made sources of optical radiation have been developed with similar or even greater potential for injury.

Acute effects on the skin are erythema, tanning and some systemic effects, chronic effects include skin aging and different kinds of skin cancer.

There are the following seven different types of injuries to eye and skin:

- Ultraviolet (UV) photochemical injury to the cornea and conjunctiva of the eye (180 to 400 nm), known as photokeratoconjunctivitis, or simply photokeratitis. This type of injury is transient and lasts for 1-2 days.
- Photochemical damage to the lens (295-325, and perhaps to 400 nm) which may produce opacities (cataracts). Although acute-exposure cataracts are possible at extremely high exposures, it is chronic exposure which accounts for environmental UV cataracts.
- Thermal injury to the retina of the eye (380 to 1400 nm). Normally this type of injury is only possible from lasers or from a very intense xenon-arc source or the nuclear fireball. The retina local burn of the retina results in a blind spot (scotoma).
- Blue-light photochemical injury of the retina of the eye (principally 380 to 550 nm; 300 to 550 nm for the aphakic eye); “blue light” photoretinitis, e.g. *solar retinitis*. Solar retinitis was once referred to as “eclipse blindness” and associated “retinal burn”. Only in recent years has it become clear that photoretinitis results from a photochemical injury mechanism following exposure of the retina to shorter wavelengths in the visible spectrum, i.e., violet and blue light. Until two decades ago, it was thought to be a thermal injury mechanism. In contrast to blue light, IR-A is very ineffective in producing retinal injuries.
- Near-infrared thermal injury of the crystalline lens (approximately 800 to 3000 nm) with potential for industrial heat cataract. The average corneal exposure from infrared radiation in sunlight is of the order of 1 mWcm^{-2} (i.e., 10 Wm^{-2}). By comparison, glass and steel workers exposed to infrared irradiances of the order of 80 to 400 mWcm^{-2} (0.8 to 4 kW m^{-2}) daily for 10-15 years have reportedly developed lenticular opacities. These spectral bands include IR-A (700-1400 nm) and IR-B (1.4-3.0 μm).
- Thermal injury (burns) of the skin (approximately 380 nm to 1 mm) and cornea (approximately 1400 nm to 1 mm; this type of injury is almost exclusively limited to laser radiation exposure); and
- Photosensitized injury of the skin, which is generally far more typical of UV wavelengths (less than 380 nm), although such photosensitized reactions can extend to approximately 700 nm, possibly as a side-effect of certain medications.

Depending upon wavelength, laser radiation can produce the same adverse biological effects and radiation of similar wavelengths from conventional optical radiation sources. The eye is normally the most vulnerable organ, and this is particularly true for laser wavelengths in the retinal-hazard spectral region (400-1400 nm). The very high radiance (“brightness”) of a laser is responsible for the laser’s significant hazard to the eye. The retinal injury is always larger because of heat flow and acoustic transients, and even a small disturbance of the retina can be significant. This is particularly important in the area of central vision referred to by eye specialists as the *macula lutea* (yellow spot), or simply the “macula”.

Outside the 400-1.400 nm retinal hazard region, the cornea –and even the lens- can be damaged by laser beam exposures. The damage mechanism in the ultraviolet spectrum is normally photochemical, whereas, in the infrared region, it is thermal in nature.

There are different sets of guidelines for broad-band optical sources and for lasers. In developing exposure limits for broad-band optical sources, action spectra were specified and used to spectrally weight source emissions to derive a “biologically effective radiance or irradiance”. This provides the most accurate hazard assessment; exposure limits can then be specified in terms of exposure duration and other relevant parameters so that all sources are evaluated with the same risk criteria.

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3.1 Introduction

Dosimetry and Diagnostic Radiology and Radiotherapy

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Only a few months after Röntgen's discovery of the X-rays first attempts were made to use the radiation for therapeutic medical applications. In the absence of any means for quantifying the 'amount' of radiation it is not surprising that it did not take too long until the first radiation caused damages became apparent. In principle, this was the birth of dosimetry. This means that a sense of awareness was created for limiting the exposure of both patients and medical personal. It does not mean that physical concepts for quantifying ionizing radiation were rapidly developed. The first sizable dose quantity appears to be the erythema dose: the amount of radiation that caused a visible reddening of the skin. Furthermore, it became quite soon evident that the appearance of an erythema also depends on the voltage chosen for the generation of the X-rays. Again the voltages themselves were not measured but estimated by examining the length over which an electrical discharge could be initiated or maintained.

In fact, it took until the year 1928 that, in the inaugurating session of the International Committee for Radiation Units and Measurement (ICRU), the first rigorously defined dosimetric quantity was developed and eventually internationally agreed upon: the exposure.

Until the late 1970s the exposure continued to be the principal dosimetric quantity for virtually all applications of photon radiation in radiation protection, diagnostic radiology and radiation therapy.

With the worldwide adaptation of the SI-system the exposure in units of Röntgens became obsolete and the attempts to extend the use of the exposure in SI units, i.e. in C/kg, were unsuccessful. As a result of this development in Report 33 [80ICR] and later in Report 60 [98ICR] the ICRU recommended for dose measurements in air the use of the quantity air kerma, where the term kerma stands for the acronym kinetic energy released per unit mass [80ICR]. As will be shown in Sect. 3.2.1.5 air kerma and exposure are intimately related quantities.

Measurements in free air serve primarily the purpose of quantifying a radiation field. The radiation field is usually only marginally modified through the measurement set up. Interactions of the radiation field with matter are not of prime concern, although interactions are necessary for generating the signal, e.g. from an ionization chamber. The situation is different when substantial amounts of matter in the beam lead to a significant modification of the radiation field. Such a situation is typical of radiation therapy, where the accurate knowledge of the dose at a given point in the real radiation field in the patient is of prime importance. Besides the primary radiation incident on the patient the real radiation field includes scattered radiation and possibly secondary particles, like e.g. recoil protons in the case of neutron therapy. Under this kind of conditions measurements are performed in-phantom in terms of the absorbed dose to water. Water is chosen as the reference material as its radiation transport properties resemble closely those of many tissues in the human body.

This chapter contains two main sections: one on dosimetry in diagnostic radiology based on air kerma and related quantities. Important documents are Report 75 of the ICRU [05ICR] and Technical Report Series No. 457 of the International Atomic Energy Agency (IAEA) [07IAE]. The other main section is on dosimetry both in external radiation therapy and in brachytherapy that is based on absorbed dose to water. Important documents are the report of Task Group 51 of the American Association of Physicists in Medicine [99AAP], Technical Report Series No. 398 of the IAEA [00IAE], Report 64 of the ICRU [01ICR] and Deutsches Institut für Normung DIN 6800 Teil 2 [08DIN].

3.1.1 References for 3.1

- 80ICR International Commission on Radiation Units and Measurements: Radiation quantities and units. ICRU Report 33, ICRU Publications, Bethesda, MD, 1980.
- 98ICR International Commission on Radiation Units and Measurements: Fundamental quantities and units, ICRU Report 60, ICRU Publications, Bethesda, MD, 1998.
- 99AAP American Association of Physicists in Medicine, Task Group 51: Protocol for clinical reference dosimetry of high-energy photon and electron beams, *Med. Phys.* **26** (1999) 1847–1870.
- 00IAE International Atomic Energy Agency: Absorbed Dose Determination in External Beam Therapy: An International code of Practice Based on Standards of Absorbed Dose to Water, IAEA Technical Report Series No. 398 Vienna, 2000
- 01ICR International Commission on Radiation Units and Measurements: Dosimetry of High Energy Photon Beams Based on Standards of Absorbed Dose to Water, ICRU Report 64, ICRU Publications, Bethesda, MD, 2001.
- 05ICR International Commission on Radiation Units and Measurements: Patient Dosimetry for X-rays used in Medical imaging. ICRU Report 74., ICRU Publications, Bethesda, MD, 2005.
- 07IAE International Atomic Energy Agency: Dosimetry in Diagnostic Radiology: An International Code of Practice, IAEA Technical Report Series No. 457 Vienna, 2007
- 08DIN Deutsches Institut für Normung: Dosismessverfahren nach der Sondenmethode für Photonen- und Elektronenstrahlung – Teil 2: Dosimetrie hochenergetischer Photonen- und Elektronenstrahlung mit Ionisationskammern, DIN 6800 Teil 2, Deutsches Institut für Normung e.V. – Beuth Verlag: Berlin 2008

3.2 Dosimetric Concepts

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For further information on dosimetric quantities see Reports 33 and 60 of the International Commission on Radiation Units and Measurement (ICRU) [80ICR, 98ICR], for radiation protection in particular Chapter 4 in Landolt-Börnstein Vol. VIII/4 "Radiological Protection" [05Kau].

3.2.1 Photon radiation

Electromagnetic radiation is the transport of energy through space. The transport does not require the presence of a medium in space, although the presence of a medium and its properties have an influence on the observable phenomena. The discrete portions of energy of electromagnetic radiation, that is also the photons of x- and gamma radiation, propagate through space with the speed of light. In contrast, charged or uncharged particles with a finite rest mass have, depending on their energy, some speed below that of light. If electromagnetic radiation propagates through a layer of material of finite thickness, a fraction of the incident photons propagates through the layer on a straight path without undergoing any interaction (transmission). The remaining part undergoes, at statistically distributed depths an interaction in which its energy and/or direction is altered (absorption/scattering).

3.2.1.1 Beam attenuation

Narrow beam

After traveling through a layer of material of thickness dx a narrow beam consisting initially of N_0 photons of a given energy will lose, through absorption and scattering, dN photons (see Fig. 3.2.1a). By integration one obtains from this the linear attenuation law:

$$N = N_0 e^{-\mu x} \quad (3.2.1)$$

where N_0 is the number of incident photons and N their number after penetration of a layer of thickness x , e is the base of the natural logarithm, and μ is the so-called linear attenuation coefficient which depends on the kind and density of the material irradiated and on the energy of the photons. It is usually given in units of cm^{-1} and is made up of the contributions for absorption and scattering

$$\mu = \mu_{\text{scatt}} + \mu_{\text{abs}} \quad (3.2.2)$$

The reciprocal value of the linear attenuation coefficient, μ^{-1} , is referred to as the mean free path length of a photon. If dN photons hit an area of size dA perpendicularly, the ratio $\phi = dN/dA$ is termed the fluence, sometimes also called particle fluence. With photons of energy $h\nu$ one obtains the ratio $\Psi = h\nu dN/dA$ which is denoted as the energy fluence. Considering the time t , in which the N photons cross the area element, one obtains the fluence rate

$$\dot{\phi} = \frac{d^2N}{dAdt}$$

or energy fluence rate

$$\dot{\Psi} = \frac{h\nu d^2N}{dAdt}$$

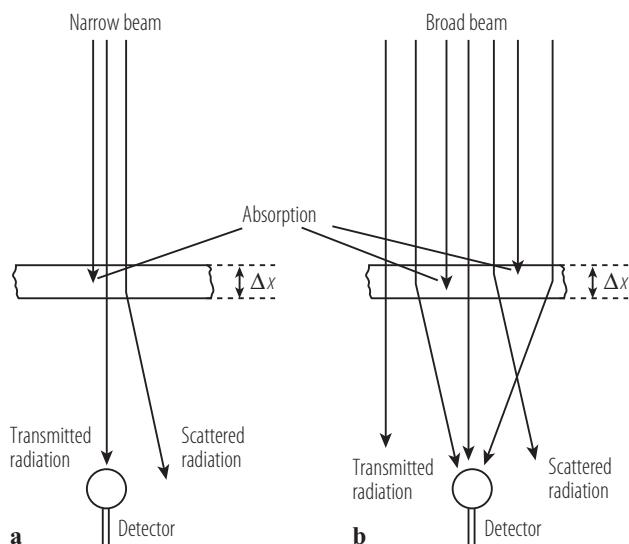


Fig. 3.2.1 In a narrow beam geometry, (a) contributions of directly transmitted radiation is recorded, while in a broad beam geometry (b) also contributions from scattered radiation are recorded.

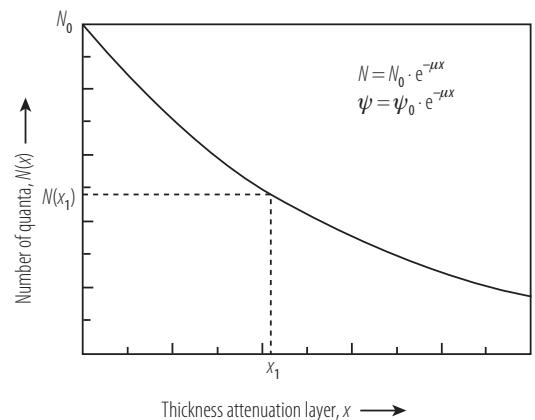


Fig. 3.2.2 In a narrow beam geometry the (energy) fluence of a monoenergetic photon beam follows an exponential attenuation after crossing a layer of material with thickness x .

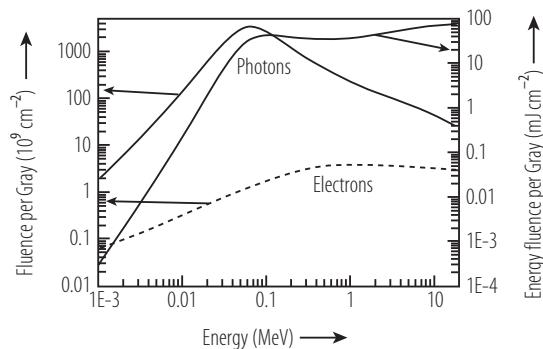


Fig. 3.2.3 Fluence (left) and energy fluence (right) of photons required to result in a small volume element of air in an air kerma of 1 Gy. Also included is the fluence of electrons to produce a dose of 1 Gy.

The fluence, fluence rate, energy fluence and energy fluence rate have units of m^{-2} , $\text{m}^{-2}\text{s}^{-1}$, J/m^2 und W/m^2 . Other frequently used units are cm^{-2} , $\text{cm}^{-2}\text{s}^{-1}$, J/cm^2 und W/cm^2 . The energy fluence is sometimes also given in units of eV/cm^2 , where one electron volt (eV) is the kinetic energy, acquired by an electron after running through a potential difference of one volt. A relation similar to that given in eq. (3.2.1) holds for the energy fluence and energy fluence rate, see Fig. 3.2.2.

$$\Psi = \Psi_0 e^{-\mu x} \quad (3.2.3)$$

That part of the incident energy fluence lost in interaction processes in a mass element in the irradiated matter produces the kerma which, in turn, finally leads to the absorbed dose defined in general terms further down in this Section and specified in greater detail in Sect. 3.2.2.3. The relation between the number of incident photons, the energy fluence and the resultant dose or kerma is shown in Fig. 3.2.3.

The linear attenuation coefficient divided by the density ρ of the attenuating material is called the mass-attenuation coefficient μ/ρ . It depends, like the linear attenuation coefficient, on the photon energy and on the atomic number Z of the material, not, however on the density of the attenuating material. This implies that the mass-attenuation coefficient of a given material is independent of whether radiation transport occurs in a gaseous, liquid or solid phase. It is the reciprocal value of the area specific mass, which is usually given in units of g/cm^2 .

The use of the area specific mass as a measure of the attenuating properties of a layer of material is convenient, as it allows an immediate comparison between different materials or identical materials in different aggregate states: an area specific mass of $1 \text{ g}/\text{cm}^2$ can be realized by an air column approximately 8 m in length, a water column of one cm length and a 1.3 mm thin plate of iron.

Broad beam

In most cases radiation fields do not fulfill the requirements for narrow beam conditions, instead they have a finite width. In measuring the attenuation caused by a layer of material in a broad beam not only photons directly transmitted are detected, but also scattered ones, as shown in Fig. 3.2.1b. As a consequence the attenuation measured in a broad beam is weaker than under narrow beam conditions. Unfortunately, there is no similar solution to eq. (3.2.1) for the general case of a broad beam falling onto a layer of material.

3.2.1.2 Interaction processes

Photon interactions are distinguished between scattering, without and with energy loss (coherent and incoherent), and absorption. For a schematic presentation see Fig. 3.2.4.

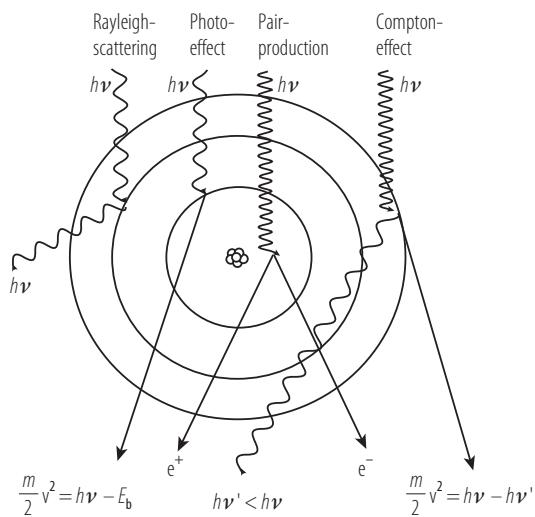


Fig. 3.2.4 Schematic presentation of different photon interaction processes

Coherent scattering processes

In Rayleigh scattering (Rayleigh scatter coefficient, μ_{coh}) a photon excites an atomic electron to oscillations with its own frequency with the consequence that the oscillating electron emits a photon of the same energy to another direction. This process leads to a change of the direction of the radiation leaving the energy, or the wavelength, unchanged. For materials with an effective atomic number between about 6 and 8 (biological tissue) and for energies above about 10 keV, that is in virtually all radiological applications, coherent scattering has no practical importance.

Scattering processes with energy loss

In Compton scattering (Compton scattering coefficient, μ_{incoh}) photons are scattered by electrons whereby they lose energy. The photon transfers a fraction of its initial energy E_{tr} to the electron. When the amount of energy is sufficiently high this results in an ionization, and a free electron with a kinetic energy $E_{\text{kin}} = E_{\text{tr}} - E_{\text{b}}$ is created, where E_{b} is the electron's binding energy. Relative to its incident direction the photon undergoes a change of the direction of propagation which is in the range between 0° und 180° and the Compton electron is emitted under an angle between 0° and 90° . The kinematics of the process are governed by the conservation of energy and momentum in the photon-electron system. The photon's scattering angle increases with increasing energy transfer. In the limiting cases of scattering angles of 0° und 180° the electron is emitted under an angle of 90° and 0° , respectively. A concise review of Compton scattering is found in the article by R.D. Evans [58Eva].

Absorption

Undergoing a photoelectric effect (photoabsorption coefficient, μ_{pe}) the photon transfers all its energy to an electron in the process of which the photon is annihilated. The resulting free electron has thus a kinetic energy $E_{\text{kin}} = h\nu - E_{\text{b}}$. For materials with an effective atomic number in the range of biological tissues ($6 < Z_{\text{eff}} < 8$) the photoelectric effect plays a particularly important role for energies below about 50 keV.

Pair Production

In a pair production process (pair production coefficient, μ_{pair}) electromagnetic energy is converted into mass. For photons with energies corresponding to at least two electron rest masses (m_e),

$$h\nu \geq 2 m_e c^2 \sim 1.022 \text{ MeV} \quad (3.2.4)$$

and in the field of the atomic nucleus this energy can be converted to a positron-electron pair. The energy exceeding the double rest mass of 1.022 MeV is transferred to the two escaping particles in the form of kinetic energy. In the later stages of this process the positron recombines with an electron which, in the case of both particles being at rest, leads to the emission two 511 keV photons in opposite directions (annihilation radiation).

At this point it should be added that there is also a nuclear photo absorption with a threshold energy typically in the range of 10 MeV. In this process the absorption of the photon leads to an excited nucleus which comes back to its ground state by emitting a neutron, (γ, n) -process, and at higher energies alternatively a proton, (γ, p) -process. In the energy range of interest here, (γ, n) -processes can be relevant in terms of radiation protection in the proximity to the radiation head of electron accelerators.

The linear attenuation coefficient μ of the attenuation law is composed of the contributions from absorption, μ_{abs} , and scattering, μ_{scatt} , and is the sum of the contributions from the individual interaction processes:

$$\mu = \mu_{\text{scatt}} + \mu_{\text{abs}} = \mu_{\text{coh}} + \mu_{\text{incoh}} + \mu_{\text{pe}} + \mu_{\text{pair}}. \quad (3.2.5)$$

In all processes involving photon energy losses considered here the energy is transferred to an electron. The electron, in turn, dissipates its initial energy along its path to other electrons in ionization and excitation processes. When by these processes an energy amount ΔE is transferred to a mass element Δm , the dose D to this mass element is obtained by $D = \Delta E / \Delta m$ (see Sect. 3.2.2.3). The magnitude of the dose is considered as an indicator of biological effects likely to occur.

Instead of linear attenuation coefficients sometimes corresponding cross sections are used. Irrespective of the kind i of the interaction process the linear attenuation coefficient and cross-section associated with process i are related to each other by:

$$\frac{\mu_i}{\rho} = \frac{\sigma_i}{u M} = \frac{N_A \sigma_i}{M}, \quad (3.2.6)$$

where σ is the cross-section per atom in cm^2 , u the atomic mass unit, N_A Avogadro's number and M the relative atomic mass of the material concerned.

3.2.1.3 Mass-energy absorption coefficient

Up to now interactions were considered just in terms of their frequency of occurrence, a view which does not allow any conclusion with respect to the amount of energy transferred. In a Rayleigh scattering process the energy transfer is zero while a photon is annihilated in a photoelectric process. In both cases an interaction has occurred, however with grossly different consequences. In order to account for the energy transferred in interaction processes the mass-energy transfer coefficient μ_{tr}/ρ and the mass-energy absorption coefficient μ_{en}/ρ are used. In simple words, the mass energy-transfer coefficient is obtained from the mass-attenuation coefficient by multiplication with the factor

$$\frac{E_{in} - E_{out}}{E_{out}}$$

where E_{in} and E_{out} are the energies of the incoming and the average energies of the outgoing photons. The mass energy-absorption coefficient is obtained by decreasing the energy in the numerator above by the amount of energy which the outgoing *electron* converts into bremsstrahlung. For many considerations, both mass-energy transfer and mass-energy absorption coefficients are extremely useful quantities. They establish, for instance, a direct link between the fluence in a photon field and the kerma (see [Sect. 3.2.1.5](#)) and are therefore frequently used in radiation transport simulations. The mass-energy absorption coefficient is also the decisive parameter governing the energy dependence of response of a detector material in cases where electron transport phenomena are not important. In other words, a detector of material m can be expected to have a response independent of energy with respect to the air kerma (see [Sect. 3.2.1.5](#)) when the mass-energy transfer coefficient of material m is similar to that of air. In this case one speaks of an air-equivalent detector.

Nevertheless a word of caution should be added here, as there may be situations in which particularly the energy absorption coefficient lacks some definitional rigidity. This is caused by the vagueness in the term 'volume of interest' in the definition of the kerma [[98ICR](#)]. As an example consider a K-shell ionization in lead. Such an event is usually followed by a cascade of de-excitation processes. As a matter of definition, all photon contributions within this cascade are disregarded in the calculation of the mass-energy absorption coefficient, as their energy is not deposited 'locally'. However, an outer-shell fluorescent photon may have a shorter mean free path than the range of the primary electron set free in the initial ionization process. For more detailed information on photon interaction cross sections, attenuation coefficients and energy absorption coefficients refer, e.g., to the review by J.H. Hubbell [[99Hub](#)].

3.2.1.4 Energy dependence of interaction processes

The contribution of the individual processes to the total attenuation coefficient depends strongly on the energy of the photon. At low, medium and high energies, photon absorption, Compton scattering and pair production are dominant, respectively. The energies where the transitions between dominance regions occur depend on the material in which the radiation propagates. This will be discussed in greater depth further down in this Section. In the energy region where the photoelectric effect is dominant the attenuation coefficient decreases steadily with increasing energy. There are discontinuous increases when the photon energy exceeds the binding energy of the next-inner shell of the absorbing atom. We talk of a so-called absorption edge, depending on the shell one speaks of K-, L-, M-edges, etc.

In Compton scattering processes the distribution of scattering angles depends on the energy of the incident photon. At low energies the angular distribution tends to be isotropic and tends to be progressively more forward directed with increasing energy. This is accompanied by an increase in energy difference between the primary and the secondary photon.

[Fig. 3.2.5](#) is a polar diagram of the angular distribution of Compton-photons. The distance between the centre and the points on a curve for a given incident energy is proportional to the probability for a scattering process with this scattering angle (differential cross-section). It is obvious that the angular distribution gets progressively more forward directed as the energy of the incident photon increases.

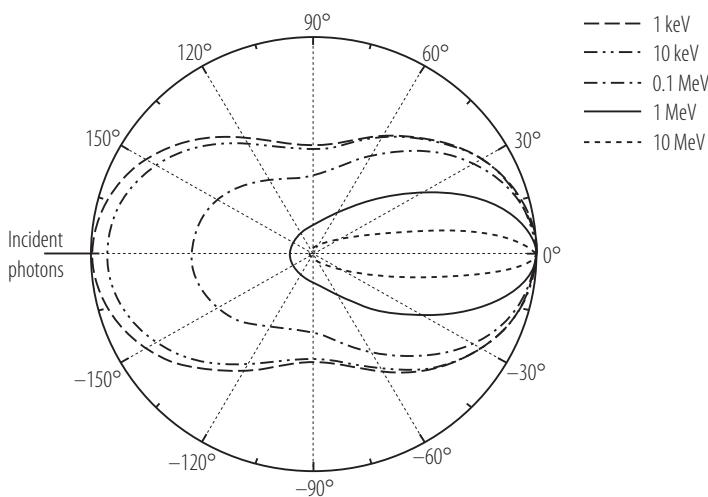


Fig. 3.2.5 Relative probability density for a Compton scattering process in which the photon is scattered by an angle θ for a selection of energies of the incident photon.

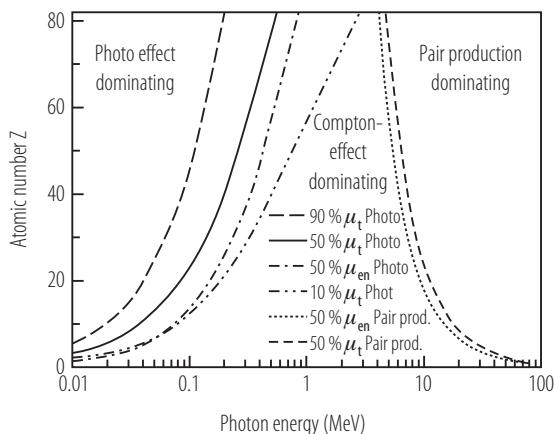


Fig. 3.2.6 Representation of the dominating kind of interaction as a function of photon energy and atomic number of the material in which the interaction takes place. For low-Z materials there is a wide energy range in which Compton scattering is the dominating process both in terms of the number of processes (μ) and in terms of the energy involved (μ_{en})

Fig. 3.2.6 gives an overview on the dominating kind of interaction process as a function of energy and of atomic number of the material in which the radiation propagates. Also, a distinction is made between the frequency of occurrence, μ_t , and the frequency of occurrence weighted by the relative energy absorption μ_{en} . Taking the example of a typical biological tissue with an atomic number of $Z = 7$ one finds from the continuous curve in **Fig. 3.2.6** that photoelectric and Compton scattering have the same frequency at an energy around 25 keV. At this energy, although identical in frequency, photoelectric interactions transfer about 25 times as much energy to electrons than the Compton effect does. In order to have, on the average, the same energy transferred in each of these two processes the energy needs to be stepped up to about 53 keV (dotted curve), an energy at which Compton scattering is strongly dominating in terms of frequency.

3.2.1.5 Kerma

The basic dosimetric quantity for measurements in diagnostic radiology is the air kerma denoted by the symbol K :

$$K = \frac{\Delta E_{tr}}{\Delta m}, \quad (3.2.7)$$

where ΔE_{tr} is the kinetic energy transferred to electrons in the mass element Δm . When the mass element consists of air we speak of the air kerma K_a . The air kerma can be measured conveniently by ionometric

methods. Each electron set free in a photon interaction produces a number of pairs of charge carriers of opposite sign which is proportional to the electron's initial energy, i.e. on the average, the production of each pair of charge carriers requires a certain amount of energy, usually denoted by the symbol W (W -value). In air and for electron energies above about 5 keV one finds $W = 33.97$ eV. This means the term ΔE_{tr} is proportional to the number of charge carriers N of one sign, which is given by $N = Q/e$,

$$K = W \frac{Q}{e} \frac{1}{m} \frac{1}{1-g} = \frac{Q}{\rho V} \left(\frac{W}{e} \right) \frac{1}{1-g}, \quad (3.2.8)$$

Usually the presentation on the right hand side is used, where ρ is the density of air, V the measurement volume and g the so-called bremsstrahlung yield. The factor $(1-g)^{-1}$ corrects the total air kerma for the bremsstrahlung part which does not contribute to the electrical charge collected. Omitting the term $(1-g)^{-1}$ in eq. (3.2.8) leads to the collision air kerma. The total air kerma is closely related to the photon fluence by:

$$K = \int E \left(\frac{\mu_{\text{tr}}(E)}{\rho} \right) \phi_E(E) dE, \quad (3.2.9)$$

where $\phi_E(E) = d\phi(E)/dE$ is the photon fluence differentiated with respect to energy, which is also referred to as the spectral distribution of the photon fluence. If in eq. (3.2.9) the term μ_{tr} is replaced by μ_{en} one obtains the collision kerma. The collision kerma divided by (W/e) is the old quantity exposure usually denoted by the symbol J , it has the dimension of charge per mass. The old unit of the exposure, the Röntgen, has the value $1 \text{ R} = 2.58 \cdot 10^{-4} \text{ C/kg}$, which nowadays is considered as the definition of the Röntgen.

For the energy range up to 1 MeV the bremsstrahlung yield has a value smaller than 0.1%. Therefore the difference between total and collision air kerma is negligible in all practical applications and no distinction will be made hereafter. The difference was introduced for reasons of scientific correctness. In eq. (3.2.7) and in the following the subscript a for denoting the air kerma is omitted for reasons of brevity.

To understand the quantity kerma one should be aware that photon radiation deposits its energy in a two-step process. In the first step the photon transmits all or a fraction of its energy to an electron in the shell of an atom or molecule whereby this is usually ionized. The electron emitted in this process takes away most of the energy transmitted by the primary photon leaving behind a positively charged ion. In the second step this electron initiates a cascade of electron-electron interactions in the course of which further ion pairs are created as schematically shown in Fig. 3.2.7. The number of the ion pairs is proportional to the kinetic energy of the electron initially liberated. It is because of this proportionality that the air kerma can be directly derived from a measurement of the amount of ionization charge.

For a correct ionometric measurement of the air kerma certain precautions are required as electron transport plays an important role in the process of the energy deposition. The use of eq. (3.2.8) requires that secondary electron equilibrium be established over the measuring volume. This term denotes a situation in which the amount of energy carried into the measuring volume by electrons entering it is equal to the amount of energy carried by electrons leaving the measuring volume. Such a situation is depicted schematically in Fig. 3.2.8 for a so-called free air chamber, a device used by metrology laboratories for realizing the unit of the air kerma for X-rays. Assuming a circular entrance aperture of diameter d the measuring volume is given by $V = \pi d^2 l$. In this kind of device the secondary electron equilibrium is established by surrounding the measuring volume with layers of air of a thickness sufficient to stop the most energetic electrons which are created in photon interactions in the solid parts of the chamber. For more information on the principle of free air chambers see, e.g., Chapter 10 in Landolt-Börnstein Vol. VIII/4 [05Kau].

For clinical measurements of the air kerma eq. (3.2.8) is never used. Instead, a dosimeter is employed which is calibrated, for the radiation qualities to be used, in terms of air kerma.

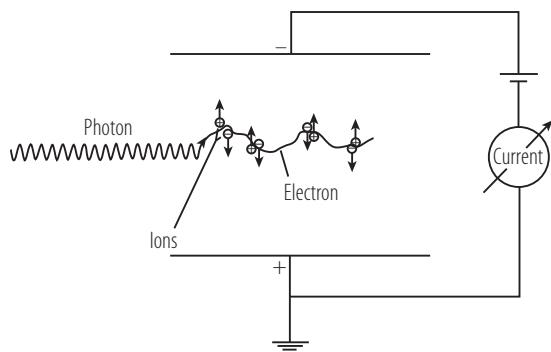


Fig. 3.2.7 In a primary interaction a photon transfers energy to an electron which, in turn produces a cascade of secondary electrons, which can be registered in an ionization chamber.

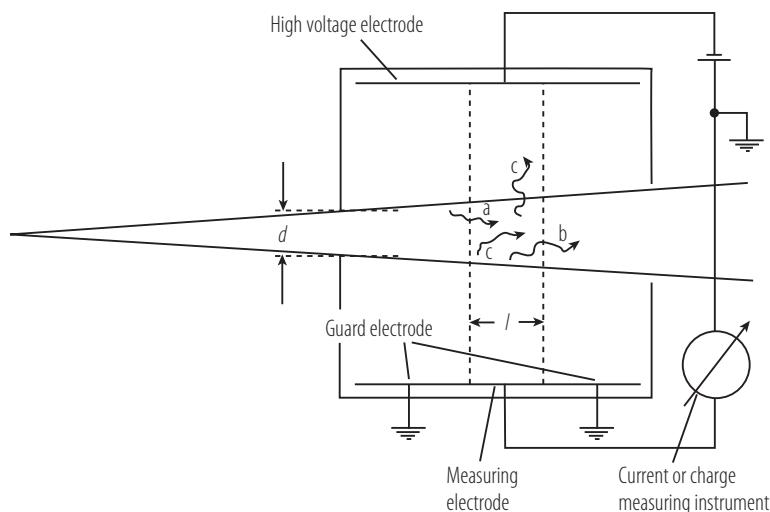


Fig. 3.2.8 Schematic presentation of a free-air chamber by means of which the air kerma can be measured according to its definition. For establishing secondary electron equilibrium the amount of energy introduced by tracks of kind *a* needs to be compensated by the energy taken away by tracks of kind *b*.

3.2.2 Electron radiation

3.2.2.1 Energetic electrons

The biological effect in radiation therapy with photon radiation is not caused by the photons themselves but by the electrons set free in photon interactions. That is why we speak of photon and also of neutron radiation as indirectly ionizing radiation. The nature of electron interactions is quite different from that of photons. While photons have a few occasional interactions from just one to may be 10 or so, electrons interact essentially with each atom they come by. Three principle kinds of interactions are distinguished: elastic interactions, inelastic collisions and radiative interactions.

3.2.2.2 Interaction processes

Elastic scattering

In elastic scattering the electron undergoes a change of direction without losing energy. The probability of elastic scattering is proportional to Z/E^2 , where Z is the atomic number of the scattering atom and E the energy of the electron.

Inelastic scattering with ionization and excitations

In inelastic scattering processes the electron undergoes also directional changes that are, however, accompanied by energy losses. A measure for the average energy loss per unit track length is the linear stopping power S :

$$S = \frac{dE}{ds} , \quad (3.2.10)$$

where dE is the energy lost over a length of track ds . The linear stopping power is usually given in units of MeV/cm. Dividing the linear stopping power by the density of the material one obtains the mass-stopping power given in units of MeV cm²/g. For a given material the latter quantity has the advantage of being independent of the density of the material considered. When needed a distinction can be made between the collision and the radiative stopping power referring to interactions in which kinetic energy is transferred to a collision partner and in which bremsstrahlung is produced, respectively.

A quantity closely related to the stopping power is the restricted stopping power S_{Δ} , defined in analogy to eq. (3.2.10) with the difference that only energy losses dE_{Δ} below a threshold Δ are considered. Energy losses above the threshold lead to electrons with energies high enough to deposit (most of) their energy outside the volume of interest, e.g., outside the sensitive volume of a detector. For dosimetry purposes the parameter Δ is usually given in units of keV. For normal thimble type chambers with a volume from 0.1 cm³ to 1.0 cm³ one often uses $\Delta=10$ keV. If there are no limitations on the value of the cut-off energy, i.e. $\Delta \rightarrow \infty$ one speaks about the unrestricted stopping power.

Bremsstrahlung production

In analogy to the stopping of electrons in the field of the atomic nuclei of anode and of target materials in accelerator heads electrons undergo stopping processes also in the nuclear field of the atoms of tissue. The energy lost by an electron when it experiences a deceleration or directional change is emitted in the form of photon radiation which is referred to as bremsstrahlung. The radiative stopping is proportional to Z^2 . For human tissues with an effective atomic number of around 7 and for energies up to 10 MeV the production of bremsstrahlung has practically no relevance. At electron energies exceeding 100 MeV bremsstrahlung production is the dominating kind of interaction before ionization and excitation.

3.2.2.3 Cavity theory, Bragg-Gray principle

Imagine a hypothetical cavity with a gaseous filling of material m inside a solid phantom of the same material. If this arrangement is exposed to electron radiation and if the cavity is sufficiently small (see further below) the dose in the gas and in the phantom close to the gas-solid interface will be equal. This is a consequence of the steadiness of the electron fluence across the interface and the spatial constancy of the fluence in the (small) cavity. As stated above, the mass-stopping power is independent of the density of the material m, which implies that the dose in the phantom close to the interface is identical to the dose in the cavity, in spite of the great difference in density. The independence of radiation transport phenomena from the density is referred to as Fano's theorem [54Fan].

Now, imagine that the gaseous filling is no longer of material m but consists of air. Again there is steadiness of the fluence across the interface and spatial constancy of the fluence in the (small) cavity. In this case the absorbed dose to material m, D_m , in the undisturbed phantom can be derived from the absorbed dose to air, D_a , in the cavity by:

$$D_m = D_a \frac{(\bar{s}/\rho)_m}{(\bar{s}/\rho)_a} , \quad (3.2.11)$$

where \bar{s}/ρ is the mass-stopping power for the materials m and air averaged over the spectral distribution of the electron fluence at the point of measurement. This is the well-known Bragg-Gray principle. If the material m is water we talk about the absorbed dose to water. This quantity is considered to be closely related to biological effects. In radiation therapy with high-energy photons, electrons and ions the Bragg-Gray principle forms the basis for dose determinations with ionometric methods.

In the description of the model outlined above it is assumed that the cavity is 'small'. In a more rigorous language the model requires the following assumptions:

- The modification of the angular and energy distribution of electrons of the first generation through the presence of the cavity is negligible.
- Over the size of the cavity, the fluence rate of electrons of all generations is constant
- The energy deposited in the cavity by secondary particles set free in interactions of the indirectly ionizing radiation within the cavity volume is negligible.

In practice, the above assumptions are never completely fulfilled. One immediately obvious example of an infringement is the necessary presence of a suitable wall around the cavity. Two approaches to deal with this problem seem particularly convenient: to choose a wall material that is either water- or air-equivalent in view of its radiation transport properties. Materials which are said to be either water- or air-equivalent never match the properties of the material to be mocked up perfectly, but usually to a good approximation. The remaining differences and other deviations from the ideal concept of perfect Bragg-Gray conditions are taken into account by so-called perturbation factors (see [Sect. 3.4.4.1.7](#))

The Bragg-Gray principle can be applied not only for the determination of the absorbed dose to water in a water phantom but also for a measurement of the air kerma in free air. If both measurements are attempted, e.g., with one suitable chamber, different perturbation factors need to be applied. The determination of perturbation factors is either done in experimental investigations or, progressively more, by radiation transport simulations by the Monte Carlo principle (in short Monte Carlo calculations). In any case, these kinds of investigations require a highly developed expertise in the perspective field.

[Eq. \(3.2.11\)](#) cannot be used as such for the determination of the absorbed dose to material m. The absorbed dose to air has to be obtained from the number of charge carriers created by the ionizing radiation according to:

$$D_a = W \frac{Q}{e} \frac{1}{\rho V}, \quad (3.2.12)$$

where Q/e is the number of charge carriers (of one sign) created, W the amount of energy required for the creation of one pair of charge carries and the denominator of the third term stands for the mass of the air in the volume of measurement.

For photon energies above about 0.5 MeV the realization of the unit of the air kerma is performed by means of ionization chambers with air equivalent chamber walls, usually graphite. The conversion of the charge measured to the air kerma relies on the Bragg-Gray principle. Instead of considering the energy imparted to the measuring volume by the primary photon radiation one looks at the energy imparted to the measuring volume by electrons which are set free by photons interacting in the chamber walls. The crucial point in this model is that the fluence and its spectral distribution of electrons do not change at the interface of two different materials. The spectral distribution of the fluence of electrons entering the air volume is given by the spectral fluence distribution of the electrons in the graphite. In this case one may apply the Bragg-Gray principle and obtains the absorbed dose to the graphite wall D_G from the absorbed dose to air in the cavity D_a by multiplying it by the ratio - graphite to air - of the mass-stopping powers averaged over the electron fluence spectrum in the interface according to [eq. \(3.2.11\)](#), where the index m now stands for graphite.

In order to determine the air (collision) kerma by means of a cavity chamber operating under Bragg-Gray conditions the chamber wall must be sufficiently thick so as to secure that the electron spectrum at the graphite-to-air interface corresponds to the electron spectrum in graphite and conditions of secondary electron equilibrium are established. Then the air (collision) kerma is obtained from the absorbed dose to graphite by multiplying it by the ratio - air to graphite - of the mass-energy absorption coefficients averaged over the energy fluence of the photon spectrum according to:

$$K_a = D_G \frac{(\bar{\mu}_{en}/\rho)_a}{(\bar{\mu}_{en}/\rho)_G} \quad (3.2.13)$$

This is the method with which metrological laboratories realize the unit of the air kerma in the energy range from 0.5 MeV to about 8 MeV.

3.2.3 References for 3.2

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3.3 Diagnostic Radiology

Dosimetry and Diagnostic Radiology and Radiotherapy

H.-M. KRAMER, B.M. MOORES, F.-E. STIEVE

3.3.1 Introduction

F.-E. STIEVE, H.-M. KRAMER

Today, diagnostic radiology is responsible for the by far largest contribution of the general public to the exposure to man-made ionizing radiation. The average effective dose (see [Section 4.5](#) in LB Vol. VIII/4) received by each individual of the general public varies somewhat from one country to another. Typical values for industrialised countries lie approximately in the range from 1.5 mSv to 2 mSv per year (see Chapter 4 of this Volume). This makes it necessary for the professions engaged in diagnostic radiology to continuously develop and apply methods for avoiding unnecessary exposures. In some form or another these methods require the determination of the dose received by the patient. In diagnostic radiology dose measurements are performed with the following objectives:

- to avoid deterministic effects in dose-intensive procedures as e.g. in interventional radiology
- to make an assessment of the risk associated with a given procedure
- to compare different methods of examinations in view of their dose requirements
- to verify conformity of a given procedure with dose reference values, see, e.g., Report 103 of the International Commission on Radiological Protection [[07ICR](#)]

The rapid advance in the frequency of the more dose-intense modalities like CT-examinations gives an additional weight to the need for dose measurements.

In a more general sense, dose measurements are a valuable means for implementing the principles of radiation protection applicable to diagnostic radiology, which are:

- Justification
- Optimisation
- The ALARA-principle (as low as reasonably achievable), see Chapter 1 in LB Vol. VIII/4

It should be noted that the fourth principle of radiation protection, i.e. the dose limitation, is not applied in diagnostic radiology at least as long as the exposure of the patient is concerned.

Dose measurements are performed in diagnostic radiology in order to estimate organ doses or the effective dose ([Section 4.5](#), LB Vol. VIII/4). Neither of these quantities are accessible through direct measurements. Instead, an estimate of these quantities is usually obtained from a dose measurement in air, either on the body surface of the patient or in the identical location, in the coordinate system of the X-ray source, in the absence of the patient/phantom. In the first case the measured dose contains the contribution of radiation backscattered from the patient/phantom, while in the second case the measured dose is representative of the undisturbed radiation field free in air. From such measurements, the organ dose or the effective dose are obtained by means of conversion coefficients which take into account the circumstances of the exposure. The most important parameters in this context are: the region of the body examined, the field size and the radiation quality, usually specified in terms of X-ray tube voltage and aluminium half-value layer (HVL).

The main objective of this Section on dosimetry in diagnostic radiology is to introduce the methods applied in clinical practice. As the measurement method is often dictated by the modality concerned, it was decided to structure this chapter by these modalities. In order to facilitate the clinical application of what is described in this chapter the more metrological aspects of dose measurements are kept at a minimum. Nevertheless, it needs to be emphasised that reliable dose measurements require the use of properly calibrated dosimeters. Properly calibrated means that the dosimeter must have a calibration

which is traceable to a primary standard, as they are operated by national metrology institutes. Furthermore, this calibration must refer to a radiation quality or a series of radiation qualities comparable to those in which measurements are planned. All the relevant information pertaining to the calibration should be given in the calibration certificate which should be considered as an integral part of the dosimeter and which needs to be supplied with it.

For deeper going information on dosimetry in diagnostic radiology, both in what metrological aspects and clinical applications are concerned, the reader is referred to Report 74 of the International Commission on Radiation Units and Measurements (ICRU) [05ICR] and to the Technical Report Series No. 457 of the International Atomic Energy Agency (IAEA) [07IAE].

3.3.2 Dose quantities

H.-M. KRAMER

In diagnostic radiology there is an appreciable versatility in the ways in which the radiation is applied to the patient. To ensure an adequate dosimetry for each modality a range of dose quantities has evolved in the course of time. As a general rule it is worth remembering that all dose measurements aiming eventually at characterizing the patient exposure are based on the air kerma (see Sect. 3.2.1.5). This is not always expressed explicitly. So sometimes use is made of the term 'entrance surface dose' when actually referring to the 'entrance surface air kerma' (see below). For some studies, e.g., for comparing the dose requirements of different examination techniques, it is often sufficient to consider just the results of an air kerma measurement. When information on the patient exposure in terms of doses to specific organs or to the whole body, i.e. effective dose of the patient, is required such information is usually obtained by applying conversion coefficients relating the result of the measurement to the desired dose quantity (organ dose or effective dose, see Section 4.5, LB Vol. VIII/4). In the majority of cases such conversion coefficients are obtained by means of Monte Carlo simulations taking into account all relevant details of the exposure. For reasons of completeness it should be mentioned here that measurements for purposes of radiation protection (of the personal) are performed in terms of the so-called operational dose quantities, i.e. ambient dose equivalent and/or personal dose equivalent (see Section 4.5, LB Vol. VIII/4).

To start this Section it is useful to introduce the term *exposure parameters* by which the settings of the X-ray tube voltage, the X-ray tube current and the exposure time are meant. A central quantity in all forms of projection radiography is the *incident air kerma*, K_i , being defined as the air kerma at a point in a plane corresponding to the entrance surface of a specific object, e.g. a patient's breast or a standard phantom. Only the radiation incident on the object and not the radiation backscattered is included in the incident air kerma. A quantity closely related is the *entrance surface air kerma*, K_e , which is derived from the incident air kerma by multiplying it with the backscatter factor, i.e. the entrance surface air kerma is measured on the surface of the body/phantom in a way that the radiation backscattered is taken into account. For specifying the dose requirement of an image receptor the air kerma incident on this device is referred to as the *image receptor incident air kerma*, which excludes scattered radiation generated in the image receptor.

It is obvious that the magnitude of the patient exposure is not solely made up by the incident or entrance air kerma but also by the size of the radiation field. The quantity kerma-area product (KAP), sometimes also called dose-area product, has been designed to take this effect into account. It is defined as the integral of the incident air kerma over the cross-sectional area of the beam in a plane perpendicular to the beam axis. The kerma-area product has two important properties: It is (a) essentially independent of the distance from the focal spot of the X-ray tube and represents (b) to a good approximation the energy imparted by the X-ray beam to the patient. Point (a) follows from the fact that, neglecting absorption and scattering processes in the air, the air kerma rate is proportional to the inverse square of the distance from the focal spot, while the field size increases with the square of the distance from the focal spot. Point (b) is explained by the fact that in most cases the majority of the radiation incident on the patient gets absorbed by the patient. Only a small fraction is actually transmitted through the patient either with no

interaction at all, which part contains the imaging information, or through single or multiple scattering events. It is essentially property (b) of the kerma-area product that makes it apt for characterizing the total exposure associated with an examination. For this reason diagnostic reference levels [99EC, 07ICR] are given in terms of the kerma-area product for a great number of examinations. In spite of the undisputed merits of the kerma-area product a word of caution appears to be adequate here. The kind and number of organs actually in the beam has, of course, an influence on the effective dose, but not on the kerma-area product. For further information on this topic, see e.g. [94Har, 04Wal], and on the methods of measurement of the kerma-area product, see Sect. 3.3.4.4.

In computed tomography the geometry of the exposure is very different from that in normal projection radiography. Therefore a group of dose quantities has been specifically designed for computed tomography. They fall under the general heading of the dose-length product which is the line integral of the air kerma along a linear section either on the system axis or parallel to the system axis. Based on this concept a group of Computed Tomography Dose Indices (CTDI) are used in practice. CTDIs are usually given in units of mGy.

The *CTDI in free air*, denoted by the symbol $\text{CTDI}_{100,A}$, is the integral of the air kerma $K_A(z)$ along the rotational axis A for one rotation of the scanner in free air divided by the product of the number of tomographic sections, n , and the nominal tomographic section thickness, T .

$$\text{CTDI}_{100,A} = \frac{1}{nT} \int_{-50\text{mm}}^{+50\text{mm}} K_A(z) dz \quad (3.3.1)$$

Here and in the following the index 100 stands for the length of integration.

The *CTDI for the CT head phantom* [09IEC], denoted by the symbol $\text{CTDI}_{100,H}$ is the integral of the air kerma $K_H(z)$ in the head phantom, H, along an axis parallel to the rotational axis of the scanner for one scanner's rotation, divided by the product of the number of tomographic sections and the nominal tomographic section thickness

$$\text{CTDI}_{100,H} = \frac{1}{nT} \int_{-50\text{mm}}^{+50\text{mm}} K_H(z) dz \quad (3.3.2)$$

The *CTDI for the CT body phantom* [09IEC], denoted by the symbol $\text{CTDI}_{100,B}$ is the integral of the air kerma $K_B(z)$ in the body phantom B along an axis parallel to the rotational axis of the scanner for one scanner's rotation, divided by the product of the number of tomographic sections and the nominal tomographic section thickness

$$\text{CTDI}_{100,B} = \frac{1}{nT} \int_{-50\text{mm}}^{+50\text{mm}} K_B(z) dz \quad (3.3.3)$$

The *weighted CTDI* for body or head, denoted by the symbols $\text{CTDI}_{w,H}$ or $\text{CTDI}_{w,B}$, are the weighted mean values of the CTDI in the phantom cross-section for one scanner rotation. The average is obtained from the CTDI-values in the centre ($\text{CTDI}_{100,c}$) and on the periphery ($\text{CTDI}_{100,p}$) of the head or body phantom.

$$\text{CTDI}_{w,H} = (1/3) \text{CTDI}_{100,H,c} + (2/3) \text{CTDI}_{100,H,p} \quad (3.3.4)$$

$$\text{CTDI}_{w,B} = (1/3) \text{CTDI}_{100,B,c} + (2/3) \text{CTDI}_{100,B,p} \quad (3.3.5)$$

The *CT pitch factor* is used in helical scanning and denotes the ratio of the patient support travel Δd along the z direction per rotation of the X-ray source divided by the product of the nominal tomographic section thickness T and the number of tomographic sections n :

$$\text{CT pitch factor} = \frac{\Delta d}{n \cdot T}$$

The volume CTDI, denoted by the symbol CTDI_{vol} is the weighted CTDI corrected for the CT pitch factor:

$$\text{CTDI}_{\text{vol},H} = \frac{\text{CTDI}_{w,H}}{p} \quad (3.3.6)$$

or

$$\text{CTDI}_{\text{vol},B} = \frac{\text{CTDI}_{w,B}}{p}. \quad (3.3.7)$$

3.3.3 Characterization of the radiation quality

B.M. MOORES, H.-M. KRAMER

The physical quantity of primary interest in dosimetry for diagnostic radiology is the dose which can be considered as the ‘amount of radiation’. Attention needs to be paid to the radiation quality. At least some information on the radiation quality is required whenever a dosimetric detector exhibits an energy dependence of its response, a situation applicable to virtually every detector, though with a wide range of severity. The quality of an X-ray beam is fully characterized by $\phi_E(E)$, its photon fluence differentiated with respect to energy:

$$\phi_E(E) = \frac{d^2N(E)}{dA \cdot dE} \quad (3.3.8)$$

where $N(E)$ is the total number of photons hitting a cross-sectional area element dA . $\phi_E(E)$ is sometimes also referred to as the spectral fluence. Fig. 3.3.1 shows a series of photon fluence spectra generated with a X-ray tube voltage of 80 kV covering the range from moderate to hard filtration. It is immediately obvious from Fig. 3.3.1 that increasing filtration (a) reduces the total fluence and (b) increases the mean energy of the spectral distribution. Effect (a) is referred to as attenuation and (b) as (beam) hardening. Consequently one may talk of attenuation equivalent filtration of two different filtering materials when both filters give the same reduction in dose rate and of beam hardening equivalent filtration when both filters lead to an identical beam hardening.

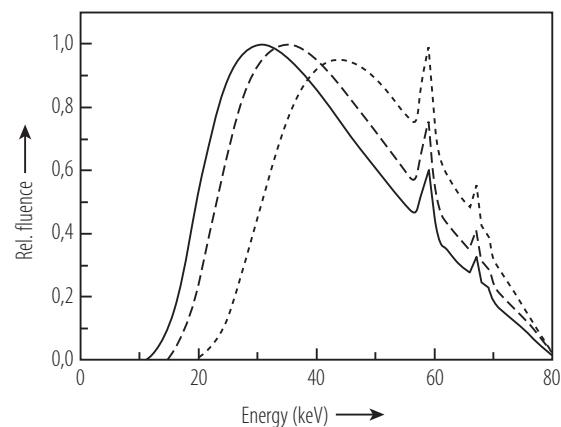


Fig. 3.3.1 X-ray spectra, 80 kV,
 (1) continuous line: 1.3 mm Al filtration, 1.84 mm Al HVL, $\bar{E} = 40.3$ keV,
 (2) broken line: 2.5 mm Al filtration, 2.73 mm Al HVL, $\bar{E} = 43.4$ keV,
 (3) dotted line: 7.2 mm Al filtration, 4.76 mm Al HVL, $\bar{E} = 48.9$ keV,
 mean energies averaged over fluence. Maximum fluence normalized to one for each spectrum. For a given X-ray tube current the spectra have the following air kerma rate ratios:
 $\dot{K}(2.5 \text{ mm Al})/\dot{K}(1.3 \text{ mm AL}) = 0.92$ and
 $\dot{K}(7.2 \text{ mm Al})/\dot{K}(1.3 \text{ mm AL}) = 0.32$.

The experimental determination of the spectral photon fluence is a demanding task which generally goes beyond the means available in a clinic. In brief it requires an energy dispersive semiconductor detector which is capable of measuring a so-called pulse-height spectrum. By means of the response matrix of the detector employed the pulse-height spectrum may be converted to the spectral photon fluence. Usually the response matrix is determined by means of Monte Carlo calculations that take into account the exact geometrical and chemical properties of the detector. For further information see references [79Bir, 79See, 00Ank]. An alternative method of obtaining photon fluence spectra is by computational means [79Bir1, 87Ile].

Because of the problems associated with a complete characterization of the photon fluence spectrum a number of parameters have been established for practical purposes that allow an approximate

characterization of the radiation quality. These are the X-ray tube voltage, the anode material and anode angle [90Kra], the total filtration being made up of the inherent filtration built into the X-ray tube and of a possibly present added filtration and the first and second half-value layer (HVL). The first HVL is defined as the thickness of material required to reduce, under narrow beam conditions (see Sect. 3.2.1.1) the air kerma of an X-ray beam to one half of its unattenuated value. The second HVL is defined as the additional thickness of material required to reduce the air kerma of an X-ray beam from one half to one quarter of its unattenuated value. For the field of diagnostic radiation qualities aluminium is usually chosen as the attenuating material. As an example one might speak of a radiation quality with a HVL of 4.76 mm Al, the value pertaining to curve 3 in Fig. 3.3.1. If the HVL is quoted without further specification it refers to the first HVL. As a consequence of the energy dependence of the interaction coefficients the attenuation of a poly-energetic radiation is accompanied by hardening of the beam. This means that the second HVL is always larger than the first one. An X-ray beam generated by a tungsten target at a constant potential of 100 kV with a total filtration of 2.5 mm aluminium has a first HVL of roughly 3.34 mm Al, a second HVL of 5.15 mm Al and a third HVL of 6.8 mm Al. Only in the case of a strictly monoenergetic radiation are first and second HVLs identical. Sometimes the so-called homogeneity coefficient is used that is defined as the ratio of the first to the second HVL. In the limit of a monoenergetic radiation the homogeneity coefficient takes the value of one which is also its upper limit. A low value of the homogeneity coefficient indicates a wide spectral distribution which is synonymous to a weakly filtered spectrum. It should be borne in mind that two spectra possessing identical 1st and 2nd HVLs or, additionally, even identical X-ray tube voltage do not need to be mutually identical. This is a consequence of the fact that these parameters do not allow a complete description of the photon fluence spectrum.

A problem in the specification of the X-ray tube voltage arises because of the so-called ripple on the high voltage which denotes the fact that the high-voltage is not constant, but oscillates around some average value. Shape, frequency and amplitude of the variation cover a wide range and depend primarily on the type of high-voltage generator. A characterization of the high-voltage by a single value in spite of the presence of ripple is often made in terms of the peak voltage or sometimes as an effective voltage. These terms lack both a proper definition and a well-defined interconnection to properties of a radiographical image. A solution to this problem was found with the so-called Practical Peak Voltage (PPV). The quantity PPV is designed in such a way that two high-voltage waveforms producing identical low-level contrast have identical PPV-values, irrespective of shape, frequency and amplitude of the ripple [98Kra, 00Yue, 08Kra]. Specifications for non-invasive high-voltage measuring instruments are found in IEC 61676 [02IEC]. Reference radiation qualities for diagnostic radiology are defined in International Standard IEC 61267 [05IEC].

3.3.4 Dosimetric equipment for diagnostic radiology

B.M. MOORES, F.-E. STIEVE

3.3.4.1 Ionization dosimeters

Suitable ionization chambers/detectors and electrometers for measuring X-ray exposure are available from a number of manufacturers and these must be matched to the user's needs for a reasonable price [92Sun]. Chambers employing air as the ionizing medium are increasingly being replaced by solid-state devices due to their compactness and robustness for field measurements. However, when an extended irradiation volume is required as for example in measurements of CTDI in Computed Tomography, air chambers are often the device of choice.

Air filled ionization chambers are usually cylindrical. However, the length to diameter ratio will depend upon the type of measurement. For CTDI measurements a thin pencil-like chamber is employed with a length much greater than diameter. For measurements on fluoroscopic units where the ionization

chamber is positioned in the bucky tray close to the input face of the image intensifier, a flat parallel-plate chamber (pancake chamber) with large diameter is employed.

Ionization based dosimeters, employing either air or solid-state devices are generally unsuitable for direct measurements on patients since they overlay the clinical detail with a deleterious image of the dosimeter. However, they are suitable for calibration of the X-ray tube and generator for subsequent use in dose measurements when known clinical exposure factors are employed or for phantom based assessment of patient doses when standard exposure factors are employed. However, modern solid-state MOSFET detectors with very small detector areas (typically 0.2 mm x 0.2 mm) have now begun to address this problem [99Pee].

For accurate dosimetry ionization based dosimeters require regular calibration against a secondary standard dosimeter that has a traceable calibration to a primary standard maintained at a National Laboratory. Such devices should be calibrated for the beam energies employed in diagnostic radiology (ranging from around 30 kV molybdenum target X-ray beams employed in mammography up to 140 kV tungsten target X-ray beams employed in CT). Similarly appropriate devices should have a constant response over the dose rate range in clinical X-ray exposure settings employed (radiographic, fluoroscopic, etc.).

Absorbed dose to tissue (muscle) is related to the air kerma by the ratio of mass-energy absorption coefficients of the tissue and air. Whilst mass-energy absorption coefficients depend upon photon energy, the ratio of coefficients averaged over the energy fluence spectrum of typical diagnostic X-ray spectra do not vary appreciably from one spectrum to another. As a consequence the ratio of averaged mass-energy absorption coefficients of tissue to air is roughly 1.05 for most of the diagnostic radiation qualities with X-ray tube voltages in the range from around 50 kV to 140 kV.

3.3.4.2 Scintillation dosimeters

Scintillation dosimeters consist of a small scintillator material roughly 2-3 mm in diameter bonded on to a fibre optic cable. The cable is optically coupled to a monitor device that measures the light output as a function of X-ray exposure to the scintillator. For support the scintillator is attached to a thin plastic disk 3-4 cm in diameter that can be positioned onto the patient surface [99Wag].

This type of detector demonstrates loss of sensitivity for angular variations greater than 30° to the normal of the support disk. However, beam energy and dose rate response is satisfactory for many diagnostic applications including fluoroscopy.

3.3.4.3 Thermoluminescent dosimeters (TLDs)

TLDs have been employed in patient's dosimetry for many years due to their relatively small size and unobtrusive presence when placed on a patient's skin during an imaging procedure. The majority of TLD phosphors are available as loose powders or in more rigid forms such as compressed or sintered pellets, extruded chips or incorporated into polytetrafluoroethylene (PTFE) disks. When employing a batch of dosimeters to undertake patient exposure measurements it is very important to ensure that the variation in response of individual TLDs within a batch is known. Manufacturers will supply batches of solid TLD whose response values are matched to within specified tolerances.

As mentioned in Sect. 3.2.1.3, the tissue equivalence of a dose detector is dictated by the ratio of the mass-energy absorption coefficients of the dosimeter material relative to tissue. Ideally the dosimeter should be "tissue equivalent" over the diagnostic X-ray energy range. This is not the case for all TLD phosphors and lithium borate is the most tissue equivalent phosphor. Also, the form of packaging of the TLD phosphor material can affect the energy dependence of response of the detector [82Wal, 99Gfi]. In the case of TLDs, where the dosimeter is "read out" a certain time after exposure, the ability to maintain the dose information stored within the device is an important consideration. Loss of information with time after exposure and before read-out is termed fading.

3.3.4.4 Kerma-area product meters

The product of the air kerma and beam area, the kerma-area product (KAP), is virtually independent of the distance from the X-ray focal spot (see Sect. 3.3.2). For a measurement of the KAP it is important that the detector completely subtends the whole of the X-ray beam. Large area ionization chambers are available that can be mounted on the X-ray tube light beam diaphragm to intercept the entire beam and integrate the dose over the whole beam area. Because their reading is independent of the distance from the X-ray tube focus, the value recorded at the diaphragm is the same as that arriving at the patient entrance surface. Kerma-area product meters should comply with IEC 60580 [00IEC].

The output from the flat parallel-plate ionization chamber that serves as the kerma-area product detector, is fed into a suitable electrometer for recording.

Such systems can provide a single measure of patient exposure for single or multiple exposures (total examination) or alternatively to both the radiographic and fluoroscopic components of complex examinations. However, such devices combine the measurement of dose and beam area into a single quantity so that accurate knowledge of either factor is unknown. More recent devices have introduced a central and separate dose recording area of one square centimetre size that provides a value of the dose at the detector. This value can then be converted to the dose to the patient if the distance to the patient's skin surface is known.

Awareness of the detector's response in respect of how the chamber is installed is important. If the chamber is mounted on an under couch X-ray tube then attenuation in the couch will affect the actual dose to the patient. Thus separate calibrations may be required for under and over couch units. Also the effects of scattered radiation and non uniformity in the dose across the X-ray beam are factors for consideration. This later effect will depend upon the distance of the chamber from scattering material whether this is the patient surface or couch. Similarly the variation in calibration with field size may need to be considered particularly for smaller field sizes when the effects of beam obliquity at the edges may have a greater effect on the overall reading.

KAP devices require calibration and in so doing the device should be exposed to an X-ray beam of known area at the chamber. Comparing the KAP reading for a known beam area against a calibrated dosimeter provides a means of calibration. However, it is extremely important to ensure that any KAP device is reading in known units. A wide variety of units is encountered in practice for example Gy cm², mGy cm², cGy cm².

3.3.4.5 Dose-length product meters

B.M. MOORES

In dose-length product measurements use is made of the various CTDIs (see Sect. 3.3.2) measured free in air or in a standard CT dosimetry phantom and meters employ a pencil ionization chamber coupled to a suitable electrometer. The diameter of the chamber dictates the accuracy with which the dose measurement can be located spatially i.e. at the iso-centre of the scanner or within a specified test phantom. Normally diameters in the region of 10 mm are employed.

The chamber normally has an active length of 100 mm with a volume of roughly 3 cm³. However, it has been recommended by the Institute of Physics and Engineering in Medicine [03IPE] that the dose integration length should depend upon the slice width. Hence, the 100 mm active length would limit such measurements to a 10 mm slice width. Equally for multi – slice CT systems also the use of a 100 mm long ionization chamber may be less appropriate than for single rotation. The appropriate length of chamber is governed by its ability to sample a high proportion of the scattered radiation that enters the measurement volume. Obviously with multi-slice CT systems the effective slice width can be quite large and CTDI values may dependent upon the number of detectors involved in data collection.

3.3.5 Measurements in diagnostic radiology

B.M. MOORES, F.-E. STIEVE

Patient dose measurements in radiography may be undertaken either directly or indirectly [93Moo]. Direct measurements are undertaken at the time of the actual X-ray exposure. Indirect measurements are undertaken after the exposure has been made, by calculating the patient dose from knowledge of the radiographic factors employed using calibration factors for the X-ray tube and generator. Alternatively, indirect measurements may involve placing a dosimeter at a point corresponding to the entrance surface of a patient and making a measurement in air when standard radiographic factors are employed in an exposure.

Both direct and indirect measurements may be undertaken on:

- physical (uniform) phantoms that attempt to mimic the overall attenuation of the X-ray beam by a patient,
- anthropomorphic phantoms that both mimic the overall attenuation of the X-ray beam by the patient as well as create an anatomical-like image,
- actual patients.

Each type of measurement generally has a different type of role and function in the overall framework of scientific support to diagnostic radiology. For example use of an anthropomorphic phantom permits many exposures to be made under standard patient but variable radiographic conditions that may exist on multiple hospital sites. In this way the variations in patient doses and image quality can be studied and/or optimised. Similarly, a physical (uniform) phantom may be used to study the magnitude of backscatter factors for variable X-ray tube voltages and X-ray field sizes.

3.3.5.1 Dosimetry in radiography

3.3.5.1.1 Direct measurements

When direct radiographic dose measurements are undertaken on physical (uniform) or anthropomorphic phantoms, generally all types of dosimetric methods outlined in Sect. 3.3.4 can be applied. The image of the dosimeter on any final image is generally not a problem since it cannot affect any diagnostic outcome.

If the dosimeter is placed in contact with the entrance surface of the phantom then the resulting measurement will provide the entrance surface air kerma, K_e , with backscatter. By multiplying the entrance surface air kerma with the ratio – tissue to air - of the mass-energy absorption coefficients (see Sect. 3.2.1.3) averaged over the energy fluence spectrum the entrance surface dose to tissue is obtained. If the dosimeter is placed in absence of the phantom or some way above the entrance surface and allowing for the distance correction of the dose rate (KAP for example) then the incident air kerma, K_i , can be measured that does not include any backscatter. However measured and calculated backscatter factor data [98Pet] can be employed to assess the amount of backscatter under defined conditions of beam quality and entrance X-ray field size. Equally if a cylindrical ionization chamber is employed and the diameter of the cylinder is a reasonable fraction of the focus to skin distance, then a distance correction may be applied to the measurement in order to correct for the diameter of the chamber. If an in-air measurement is made without a patient or phantom, then care must be taken to ensure that scatter from the tabletop is minimal.

Direct radiographic dose measurements made on phantoms will provide information on patient exposure under the prescribed exposure conditions defined by the particular phantom employed. Thus if such exposures are meant to mimic the clinical situation then it is important to verify that the resulting image density (optical density for film, or exposure index for digital recording medium) is representative of that normally encountered clinically. Equally when attempting to compare measurements made with or without backscatter it is important to have detailed information on the X-ray beam conditions (X-ray tube

voltage, filtration) as well as the field sizes employed. This latter information is extremely important if KAP measurements are to be compared to entrance surface dose measurements.

When direct radiographic dose measurements are made on actual patients [07Har], it is important to employ a dosimeter that does not interfere with the clinical image. In many cases this can be achieved by positioning the dosimeter in a diagnostically insensitive area (for example, left or right shoulder for chest/lung examinations). TLD or very small MOSFET dosimeters are generally patient compatible for direct entrance surface air kerma measurements that include backscatter. KAP measurements, which do not include backscatter, do not interfere with the clinical image but it is very important to ensure that the X-ray beam has been correctly collimated. Over-collimated field sizes (too small) are noticeable on the final image but under-collimated (too large) are not.

KAP meters that provide separate measurements of the air kerma as well as kerma-area product have to be corrected for the actual focus to skin distance employed since measurements are made at the collimator output. Such measurements may also have to be corrected for backscatter.

All direct radiographic dose measurements, other than KAP, tend to be labour intensive. If phantoms are employed they have to be positioned in the X-ray beam together with the appropriate dosimeter. This type of measurement is usually the domain of the medical physicist with very little involvement of the radiographer/technician.

3.3.5.1.2 Indirect measurements

Indirect radiographic patient dose measurements employ the calibration factors for each particular X-ray tube and generator employed clinically. These consist of measured X-ray outputs undertaken at one metre or 70 cm (both bear a roughly 2:1 relationship due to the inverse square law) when a sample of the full range of clinically employed exposure factors is selected [05IPE]. For example a X-ray tube voltage range of 60-120 kV in 10 kV steps, a X-ray tube current range 100-500 mA in 100 mA steps and exposure time range of 0.1-1.0 s in 0.1 s steps. Usually a calibrated ionization type dosimeter is employed since these are robust for field use. These devices now may also employ a radio link between the dose data display unit and any PC employed for data logging into structured templates.

From a knowledge of the exposure factors employed during actual radiographic examinations it is possible to calculate the Entrance Surface Dose (ESD) to tissue by the expression [07Tsa]:

$$\text{ESD} = K_e \left(\frac{\bar{\mu}_{en}}{\rho} \right)_a^t \left(\frac{d_0}{d} \right)^2 B \quad (3.3.9)$$

Where K_e is the entrance surface air kerma measured during the tube and generator calibration,

$$\left(\frac{\bar{\mu}_{en}}{\rho} \right)_a^t$$

is the ratio – tissue, t, to air, a, - of the mass-energy absorption coefficients averaged over the energy fluence spectrum, d_0 is the focus to detector distance employed in the calibration, d is the actual focus to skin distance during a radiographic exposure and B is the backscatter factor for the X-ray tube voltage, filtration and X-ray field size employed.

Software packages that can undertake this type of calibration are available or can easily be produced. If calibration of the generator is undertaken at a number of fixed X-ray tube voltages then interpolation can be used to calculate X-ray outputs at intermediate values. Similarly if a fixed value of patient thickness is employed for all measurements of K_e for a particular examination on a specific patient group, e.g. male abdomens, then the overall uncertainty on the dose measurements can be calculated from a knowledge of the distribution of patient thicknesses encountered in practice. Alternatively a tube-mounted tape measure can provide this information.

Manual data entry of radiographic examination details into dose calculating software can be tedious and prone to error. However, direct access to radiographic exposure data that has already been entered into a Radiology Information System (RIS) provides easy access to very large amounts of patient dose data [10Moo]. Equally examination details stored on the DICOM header of each radiographic image can

also be accessed directly for patient dose evaluations. A similar approach has been developed by the manufacturers of X-ray equipment by employing exposure parameters directly from the generator together with pre-programmed calibration data to calculate patient dose. Although equally applicable to radiographic patient dose assessments such developments have tended to be applied for high dose fluoroscopic examinations and will be discussed in [Sect. 3.3.5.6](#).

Many studies have been undertaken of patient doses resulting from radiographic exposures for all of the main types of examinations and European Dose Reference Levels (DRLs) [EC96] exist for them in terms of both ESD and KAP.

3.3.5.2 Dosimetry in fluoroscopy

During fluoroscopic examinations the X-ray beam is usually moved relative to the patient, unlike radiography when the X-ray beam and patient remain in fixed relative positions. For the simpler types of examinations such as barium enema, the X-ray beam may be moved spatially relative to the patient. For more complex examinations, such as cardiological ones, the X-ray beam may move spatially but also angularly as different projections are employed. It is also possible that the field size may be varied throughout an investigation. Consequently spatial variations in dose create difficulties when trying to define realistic measures of overall risk.

Attempts to employ dosimeters (for example TLDs) fixed to the patient's surface mean that only localised samples of the entrance surface air kerma can be taken. Also changes in X-ray beam angulation as well as changes in field size can lead to individual dosimeters not being irradiated or even irradiated sub optimally along their edge [97Boo]. Hence for fluoroscopic examinations, dose assessment usually involves an integral over time of the total dose employed. This is assessed most easily by means of a kerma-area product meter or calibrated exposure factors, where the calibration involves the dose-rate at different settings. By employing a running time dependent assessment of the fluoroscopic factors used, a time integral of dose rate can be formed within the generator to establish the overall dose delivered to the patient.

For standard fluoroscopy it is possible to employ a phantom in a direct dose assessment [93Sul, 98Ver]. Reproduction of the exposure profile employed for such examinations can provide an estimate of a standard examination dose. In this context standard means examinations performed with average field size, from a particular direction involving a particular area of anatomy. For dose assessments involving actual patients direct measurements are provided by the kerma-area product [98Min]. However, running time dependent dose assessments by the generator may also be considered to be direct since the dose is calculated during the exposure.

Both approaches create information gaps when attempting to assess risks. The kerma-area product approach combines the dose and area into a single parameter. The dose component provides a measure of risk for a given set of irradiated organs and the area component determines those organs that are irradiated for a given anatomical location, including beam direction. The generator derived dose estimate provides an overall estimate of the dose delivered but without knowledge of the field size and anatomical location, therefore the specific organs irradiated are not known accurately. Obviously utilisation of the imaging information can provide details of the volumes of tissue, their location and the organs involved. But combining these two sets of data can only be undertaken on a case-by-case basis and at present is time consuming so that usually general irradiation conditions (projection, volume of tissue etc.) are assumed when assessing risks.

When both radiographic and fluoroscopic exposures are employed in a single examination both approaches can keep a record of each type of dose separately [98Gfi]. However, doses delivered during most fluoroscopic examinations are very much operator as well as case dependent. Because they are very much dependent upon the fluoroscopic screening time, this parameter is often employed as a risk related performance parameter. For extremely high dose fluoroscopic examinations such as interventional techniques, the area of the entrance surface (skin) that receives the largest accumulated dose is very relevant since deterministic effects (see Section 2.3, LB Vol. VIII/4) are possible. Consequently a localised ESD value is then of some importance. This will be discussed in [Sect. 3.3.5.6](#).

3.3.5.3 Dosimetry in mammography

3.3.5.3.1 Phantom measurements

Practical dosimetry in mammography employs a standard breast model that assumes a 5.3 cm thick breast with a glandularity of 29% in the central region with 0.5 cm thick adipose layers at the top and bottom. The composition of this model has been found to be typical for breasts of 5.3 cm compressed thickness for women in the age range 50-64. For practical dosimetry this standard breast is equivalent to a 4.5 cm thick Perspex phantom. The dosimetric quantity employed in mammography is the mean glandular dose given by Dance [00Dan]:

$$D_{\text{mgd}} = K_e \cdot g \cdot c \cdot s, \quad (3.3.10)$$

where, K_e is the entrance air kerma for a 4.5 cm thick Perspex phantom, g converts the incident air kerma for the standard 5.3 cm breast to mean glandular dose, c is the correction factor that allows for the glandularity of the standard 5.3 cm breast and s is a spectral correction factor that takes into consideration the use of different X-ray spectra produced by the use of different anode/filter and tube voltage combinations [00Dan1]. Both the g and c factors are dependent upon the half-value layer (HVL) of the X-ray beam and have been tabulated for different breast thicknesses and HVLs as well as breast composition [00Dan1].

In order to assess the mean glandular dose the Perspex phantom is exposed using the settings used in clinical practice for a breast of size and composition similar to a 5.3 cm thick standard breast. The mAs per exposure is determined by using the Perspex phantom positioned as for a crano-caudal exposure with the compression device in place against its top surface. For this mAs value the air kerma is measured directly in the plane equivalent to the top surface of the breast (Perspex) phantom but in the absence of the phantom. Alternatively this can be measured from the knowledge of the calibration of the tube and generator (incident air kerma per mAs at 50 cm) at the appropriate beam quality and mAs exposure corrected for the actual focus to breast surface distance. The effect of backscatter is not required in this measurement since it is accounted for in the calculations of the g -values that are employed in the glandular dose calculation presented above. The measured incident air kerma values are then employed in the expression for D_{mgd} with the appropriate g , c and s values for the beam quality (X-ray tube voltage, filtration).

3.3.5.3.2 Patient measurements

Periodic measurement of the mean glandular doses delivered to actual patients is recommended. For this compressed breast thickness, X-ray tube voltage and mAs delivered are recorded. From the calibration of the tube and generator for the X-ray tube voltage, target material and filtration used the mAs can be used to estimate the mean glandular dose (indirect method) from eq. (3.3.10) for the appropriate breast thickness using tabulated values of g , c and s and assuming a 50% glandularity.

3.3.5.4 Dosimetry in computed tomography

The population exposure to ionizing radiation due to medical procedures continues to rise mainly due to the increased use of imaging procedures. A major part of this increase over the past 20 years has been due to the increased use of computed tomography (CT) and the emergence of volume scanning procedures. This has mainly been driven by the evolution of multi-slice CT scanners. With this ongoing increase in utilisation of CT the dose to patients is an important consideration. The choice of dosimetry parameter and its measurement is dependent upon the purpose for which it is required.

3.3.5.4.1 Methods of measurement of CTDI

The ionization chamber is aligned along the z-axis of the scanner and irradiated by a single axial scan at the centre of the chamber length. The dosimeter should comply with IEC 61674 [97IEC]. Because it is calibrated with the whole length irradiated the value of dose recorded for the partial irradiation by a single scan will need to be corrected for the dose reading that would result from complete irradiation of the chamber. The computed tomography dose index CTDI_{100} is then given by:

$$\text{CTDI}_{100} = M \cdot N \cdot \frac{L}{n \cdot T} \quad (3.3.11)$$

N: calibration factor for the ionization chamber at appropriate energy

M: dosimeter reading

L: length of ionization chamber (100 mm)

n: number of slices acquired simultaneously

T: nominal slice thickness, or detector group width for multi-slice scanning

The term L/nT corrects the partial chamber irradiated reading *M* to the complete chamber irradiation value. The CTDI_{100} value in mGy for a given mAs is often normalised to ${}_{\text{n}}\text{CTDI}_{100}$ in mGy/mAs.

A simple measure of tube output is given by $\text{CTDI}_{100,\text{A}}$ measured in air with the ionization chamber aligned along the scanner axis positioned at the iso-centre. The CTDI can also be measured in any phantom with a suitable cavity for the chamber, normally 16 or 32 cm diameter Perspex, so-called head and body phantoms. The quantity is then termed $\text{CTDI}_{100,\text{H}}$ or $\text{CTDI}_{100,\text{B}}$. Measurements are made both at the centre and at the periphery. However, because of beam hardening effects and shaped pre-filters the dose measurements at the centre and periphery will be different. The central and periphery values can be combined to provide the weighted CTDI_w (see Sect. 3.3.2).

The CTDI_w is employed in the EC document “Quality criteria and reference doses in CT” [96EC1] and represents a single axial scan measurement. For spiral scanning CTDI can be considered to represent the dose for a single rotation of the X-ray tube in spiral mode. When spiral movements involve a pitch that is not equal to unity then the CT pitch factor is required since a given slice of tissue is not irradiated throughout a single tube rotation. The dose is then smeared out along the z-axis length. Dividing the weighted CTDI_w by the pitch factor (see Sect. 3.3.2) gives the volume CTDI_{vol} , see eqs. (3.3.6) and (3.3.7).

3.3.5.4.2 Dose length product (DLP)

The dose length product (DLP) is an indicator of the overall exposure for a complete CT examination given by the average dose multiplied by the scan length. Thus DLP is given by:

$$\text{DLP} = {}_{\text{n}}\text{CTDI}_w \cdot W \cdot n \cdot T, \quad (3.3.12)$$

where

${}_{\text{n}}\text{CTDI}_w$: weighted CTDI normalised to tube loading

W: tube loading per single slice (mAs)

n: total number of slices imaged

T: nominal slice thickness or detector group in multi-slice scanning

The above definition can also be generalised for spiral scanning.

3.3.5.4.3 The relevance of CTDI

The CTDI employing a 100 mm long pencil ionization chamber with homogeneous cylindrical Perspex dosimetry phantoms was developed in the era of single slice scanners with collimated slice thicknesses of

10 mm or less. With the advent of multi-slice CT scanners with wide (40 mm) collimated X-ray beams the efficiency and relevance of this dose measure has been questioned [06Dix].

Such discussion has centred on the role of CT dose measurements as either an index of relative scanner performance or as an index of risk to patients when organ doses are of importance.

It has been suggested that:

- Multiple organ dose measurements in an anthropomorphic phantom with a set of MOSFET dose meters are as easy to perform as CTDI measurements.
- Organ dose measurements are equally as effective for machine QA as CTDI.
- The CTDI concept has had to be continuously modified to accommodate changing CT technology.
- Organ dose measurements provide direct assessment of organ risks.

More recently attempts have been made to introduce the concept of kerma-air product (KAP) into CT in order to unify CT dosimetry with radiography and fluoroscopy. The CT KAP is given by [08Hud]:

$$KAP_{CT} = \sum_{\text{proj area}} \int K(x, y) dx dy \quad (3.3.13)$$

where $K(x,y)$ is the air kerma at a point (x,y) in a plane perpendicular to the X-ray beam axis for projection p. The integral in eq. (3.3.13) is taken over the X-ray beam area and the KAP_{CT} is obtained by summing over all projections. This approach treats the CT-exposure as being made up of a large number of radiographic projections taken from different angular directions. Such an approach would bring CT patient dosimetry into line with other X-ray modalities such as radiography and fluoroscopy but also other 3D imaging procedures such as tomosynthesis.

3.3.5.5 Dosimetry in dental radiography

There are three types of dental X-ray techniques:

- Intra-oral employing plain film or digital detector,
- Panoramic tomography,
- Dental CT

3.3.5.5.1 Intra-oral radiography

Assessment of patient doses from intra-oral dental radiography is straightforward and consists of the assessment of the entrance surface kerma for a particular view. This may be achieved by measuring the air kerma at the end of the spacer cone by placing a suitable dosimeter in the beam at a known focus-to-dosemeter distance. The detector should be small enough to lie within the X-ray beam area given the relatively small beam size employed in dental radiography. Otherwise any reading would need to be corrected for the actual volume of detector irradiated relative to the total detector volume. The dosimeter is exposed at the normal settings employed for particular dental examinations. On many sets these are pre-programmed settings. On older sets they can be manually selected in terms of exposure time for a fixed X-ray tube current. Dosemeters with solid-state detectors are the most practical due to their relatively small size.

In many studies of patient doses from dental radiography calibrated film has been employed as the dosimeter particularly when undertaking large-scale surveys [85Pan]. Film packs are sent to the dental surgery from a central laboratory and the dental practitioner exposes these at the appropriate machine settings. Alternatively TLDs may be similarly employed. All of these measurements provide a measure of dose in air without backscatter.

3.3.5.5.2 Panoramic tomography

In this technique an image of the whole dentition may be obtained on a single film or digital detector. This involves rotation of the tube and imaging device around the patient's head. However, because of the non-spherical nature of the mandible, the local centre of rotation of tube and imaging device moves to different locations within the mandible throughout a complete rotation of the skull. Normally three such localised centres of rotation are employed one each for the left, centre and right hand side of the mandible. This means that the "tomographic axis" is different for different parts of the motion. Also, the X-ray beam is collimated by aligned slits at both the tube and image receptor and a curved cassette moves across the slit collimator at the image receptor. Consequently as with all examinations involving different projections, the entrance surface kerma is not a suitable measure of patient dose.

A simple measure of dose performance involves measurement of the air kerma at the image slit collimator for a full rotation with the detector completely irradiated. This requires a detector that is small enough to be fully covered by the collimated beam. Solid-state detectors are the preferred choice. The X-ray beam width is measured by taping a film or phosphorescent plate to the front of the image collimator and exposing it throughout a full tomographic movement. From the resulting exposure image the X-ray beam width is assessed and the dose-width product can be measured for a known beam length [06Doy]. When using a phosphorescent plate this can be viewed in a darkroom following exposure.

A more comprehensive assessment of patient dose during dental ortho pantomography can be undertaken by employing an anthropomorphic head phantom (for example Rando, Alderson Research Laboratories Inc.). TLDs are placed in the phantom in anatomically relevant positions and calibration and conversion coefficients applied to give organ dose or the effective dose [79Wal, 00Gor]. Radiographic factors for each tomographic programme are selected to be appropriate for the phantom. Mean organ doses can be calculated by averaging values of TLDs placed in different positions of the same organ. If required the effective dose may be calculated by applying the appropriate tissue weighting factors [98Lec].

3.3.5.5.3 Dental CT

Dental CT units are now being employed in dental radiography. These employ either spiral CT or cone beam CT techniques. Dosimetric methods are still under development but methods under assessment are similar to those employed in general CT practices. Thus n_{CTDI_w} (see Sect. 3.3.5.4.2) or DLP may be applied where measurements are made with a 16 cm diameter head phantom.

It has been proposed that an estimate of the effective dose may be made from values of DLP by means of the expression:

$$E = f \cdot n_{CTDI_w} \quad (3.3.14)$$

where f for the head is $0.0021 \text{ mSv} \cdot \text{mGy}^{-1} \cdot \text{cm}^{-1}$ as presented in the EC Quality criteria document for CT [04EC].

3.3.5.6 Dosimetry for interventional procedures

Interventional procedures involve X-ray guided therapeutic interventions employing fluoroscopy, fluorography or CT. Some of these procedures can be complicated and involve very high doses to the patient. Also such procedures may involve spatial and angular movement of the X-ray beam with their associated problems for meaningful assessment of overall and/or local patient dose. Localised doses may be high enough to produce deterministic effects ranging from temporary epilation of the head to more serious skin injuries. For this reason patient dose assessment may need to involve not only knowledge of the overall dose delivered to the patient but also the highest local entrance surface dose.

3.3.5.6.1 Overall dose to the patient

Protocols for patient dosimetry in interventional radiology recommend the use of kerma-area product monitoring [95Van]. However, for this approach it may be difficult to relate KAP measurements to the local maximum absorbed dose to skin. Nonetheless, for procedures involving well-defined projections an array of suitably positioned TLDs on the skin of the patient may be employed to provide local and overall dose assessments [00Han, 00Moo].

Because high doses in interventional procedures often result from the use of long fluoroscopic screening times, dose evaluations in terms of maximum permitted dose rates have also been undertaken [98Gel]. Such studies can employ standard polymethylmethacrylate (PMMA) phantoms with a suitable dosimeter positioned on its surface in order to replicate the clinical procedure under standard conditions. Maximum permitted dose rates can then be defined for both the entrance surface of the patient as well as that to the image intensifier input face.

3.3.5.6.2 Maximum entrance surface dose

Both TLD and film dosimetry have been employed in the assessment of the maximum entrance surface dose for interventional procedures. A slow film such as Kodak X-Omat V film of the type employed in radiotherapy applications, is positioned at appropriate locations on the patient's entrance surface. The film is calibrated for the radiation qualities employed in the interventional procedure. Following exposure the film density is measured and dose assessed from the sensitometric curve (see Fig. 3.4.5 in Sect. 3.4.3) of the film [01Van] (see also Sect. 3.4.3).

For high dose fluoroscopic examinations real time monitoring of skin dose is sometimes performed by placing small sensors on the skin surface [99Wag]. Both metal oxide field effect transistors and scintillation dosimeters have been proposed. An alternative indirect approach is to monitor electronically the X-ray tube current, X-ray tube voltage and pulse width as well as the focus to skin distance (PEMNED, clinical Microsystems Inc, Arlington, Virginia). In this way calculations are made to monitor the dose and dose-rate to the patient using previously made tube calibration measurements. This approach will always overestimate the maximum skin dose since it cannot predict where the X-ray beam is aimed. An alternative indirect approach positions a dosimeter in the X-ray tube port.

3.3.6 References for 3.3

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3.4 Radiotherapy

Dosimetry and Diagnostic Radiology and Radiotherapy

H.-M. KRAMER, H.-J. SELBACH, S. VATNITSKY

3.4.1 Introduction

H.-M. KRAMER

The central task in radiation therapy is to prescribe and deliver the ‘correct’ dose to the patient. The term ‘correct’ is to denote that the tumor receives a sufficiently high dose so as to terminate its existence and keeping at the same time the danger of its reoccurrence at an acceptable level. Simultaneously, excessive dose levels to the organs in the tumor’s neighborhood must be avoided in order not to put their functionality at an unacceptable risk. Such an optimization can only be performed adequately by conducting suitable dose measurements. The International Commission on Radiation Units and Measurements (ICRU) has recommended already in 1976 [[76ICR](#)] that the dose to the tumor should be known with a relative standard uncertainty not exceeding 2.5%. Even by today’s standards this goal is not easy to achieve.

In essence, the dosimetry for both external radiotherapy and brachytherapy is based on characterising the output from the radiation source prior to the exposure of the patient. On the basis of this data the patient is exposed either for a predetermined time or a predetermined number of monitor units. If at all, on-patient dosimetry is performed in radiotherapy only for QA or research purposes. In the great majority of cases of external radiation therapy linear electron accelerators are employed delivering either a megavolt bremsstrahlung or an mega electron volt electron beam. The dosimetry for this form of treatment is described in [Sects. 3.4.4.1](#) and [3.4.4.2](#) which is followed by a brief section on some special treatment forms with linear electron accelerators and with radiation from X-ray tubes. Over recent years a significant increase in hadron therapy has emerged for which beams of protons or heavier ions up to and including Ne and with particle energies up to about 200 MeV/u are used. The dosimetry for this form of therapy will be dealt with in [Sect. 3.4.4.4](#). [Sect. 3.4.5](#) will be devoted to the various forms of brachytherapy.

3.4.2 Dose quantities

H.-M. KRAMER

The dosimetric quantity of interest in radiotherapy is the absorbed dose to water. The majority of dosimetry protocols, at least those for teletherapy [[96IPE2](#), [99AAP](#), [00IAE](#), [03IPE](#), [08DIN](#), [08NCS](#)], follow nowadays the concept of what could be called a native absorbed dose to water approach. This means that the complete dosimetric chain is based on reference dosimeters calibrated in terms of absorbed dose to water usually in the radiation of a ^{60}Co -source. Some older dosimetry protocols are based on reference dosimeters calibrated in the quantities air kerma or exposure (see [Sect. 3.2.1.5](#)). As also in these cases the absorbed dose to water is eventually the quantity of interest it is the task of the medical physicist in the clinic to perform the transition from air kerma/exposure to absorbed dose to water. As calibrations of reference dosimeters in terms of absorbed dose to water are available from a great number of metrology institutes and from reference dosimetry laboratories around the world the following Sections employ the concept of a native absorbed dose to water approach. The reader can find an example of a code of practice based on an air kerma calibrated reference chambers e.g. in TRS 277 [[87IAE](#)].

3.4.2.1 Photon beams

Fig. 3.4.1 is graphical presentation of the variation of the dose as a function of depth along the beam axis inside a water phantom, usually referred to as a depth dose curve. With decreasing slope the individual curves refer to beams with nominal accelerating potentials of 4, 6, 8, 10, 15 and 25 MV. All curves are normalized to one for an in-phantom depth of 10 cm. The beam crossing the phantom surface consists primarily of photons and only of a few electrons. Through interactions of photons with the water molecules more and more secondary electrons are generated leading to the so-called build-up until the dose reaches its maximum value at a depth usually denoted by the symbol z_{\max} . With further increasing depth the dose (rate) decreases monotonically without reaching the value zero. At a depth just behind z_{\max} where the curves begin to have a positive curvature and beyond secondary electron equilibrium is established.

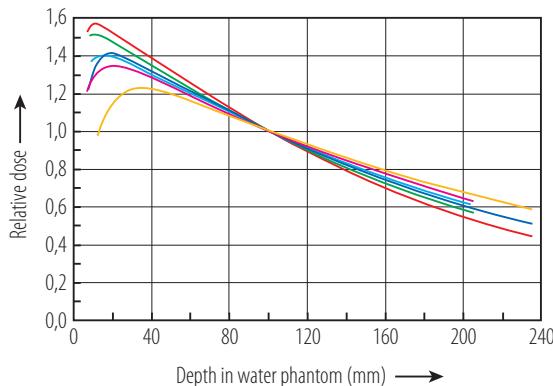


Fig. 3.4.1 Examples of depth dose curves for high-energy photon radiation with generating potentials of 4, 6, 8, 10, 15 and 25 MV from top to bottom at the left.

As the response of any dosimetric detector depends in some form on the energy of the radiation to which it is exposed there is the need for an experimental characterization of the radiation quality. Ideally, one would like to be in possession of the spectral distribution of the photon fluence. However, because of the high photon flux densities encountered in therapeutic beams there are substantial problems associated with an experimental determination of this kind of information. Therefore one uses easily accessible parameters which characterise the radiation in a suitable way e.g. in terms of its penetrating power.

One such quantity applicable to photon radiation is the percentage depth dose (PDD) defined as the ratio of the doses at depths z and z_{\max} . Usually the depth is chosen to be $z = 10$ cm in which case the symbol PDD_{10} is used. A related quantity is the tissue phantom ratio (TPR) defined as the ratio of doses in depths z_2 and z_1 inside a phantom at constant source-to-detector distance (SDD):

$$TPR = \frac{D(z_2)}{D(z_1)} \quad (\text{SDD} = \text{const.}) \cdot \quad (3.4.1)$$

Usually the depth z_2 is greater than z_1 . The TPR can be measured particularly easy in a combination of a beam directed vertically and a water phantom. Filling the water phantom to different levels without moving the detector results in different values of z at constant SDD. The special combination of $z_1 = 10$ cm and $z_2 = 20$ cm is referred to as $TPR_{20,10}$. When the field size in the plane of measurement is $10 \text{ cm} \times 10 \text{ cm}$ $TPR_{20,10}$ is called the radiation quality index Q , $Q = TPR_{20,10}$. For a number of typical radiation qualities values of Q are given in [Table 3.4.1](#).

Table 3.4.1. Approximate relation between the accelerating voltage U , in MV, and Q ; in column two Q is also given for ^{60}Co radiation

U [MV]	^{60}Co	2	4	5	6	8	10	12	15	18	21	25	35	50
Q	0.57	0.52	0.62	0.64	0.68	0.71	0.73	0.75	0.76	0.77	0.79	0.80	0.81	0.83

The accelerating voltages of a medical linear accelerator are nominal values which may differ to some extent from the actual accelerating voltage. A more realistic value of the effective accelerating voltage U_{eff} can be obtained from the radiation quality index by [65Har]:

$$U_{\text{eff}} = \exp\{7.2 - 24.9(Q - Q^2)\} \quad (3.4.2)$$

where U_{eff} is in units of MV. For a typical filtration in a medical accelerator the numerical value of U_{eff} corresponds to the average electron energy in MeV hitting the target.

When z_1 is chosen to be z_{max} , i.e. the depth of the dose maximum, the TPR has the special name tissue maximum ratio (TMR).

While the shape of a depth dose curve and hence also the value of PDD_{10} is influenced by the source-to-surface distance (SSD) the tissue-phantom ratio and also the tissue-maximum ratio is essentially SSD independent.

In cases where an ionization chamber can be moved by a computer controlled 3D positioning device inside a water phantom it may be more convenient to determine the depth dose curve, at constant SSD, rather than the tissue phantom ratio. The ionization chamber readings at depths of 10 cm and 20 cm, M_{10} and M_{20} may be used to calculate the radiation quality index according to [10DIN]:

$$Q = 1.2661 \cdot \frac{M_{20}}{M_{10}} - 0.0595 \quad (3.4.3)$$

3.4.2.2 Electron beams

Fig. 3.4.2 shows a schematic presentation of a typical depth dose curve of a high-energy electron beam. Like in the case of photon radiation there is an initial dose increase, followed by a maximum and a region of linear decrease. The initial increase is caused by an increase in electron fluence with increasing depth. Neglecting the (small) divergence of the electron beam the direction of the electrons reaching the phantom surface is essentially perpendicular to the phantom surface. Due to interactions of the electrons with the water molecules the average angle relative to the phantom entrance plane becomes less than 90° , an effect which does not change the so-called planar fluence, i.e. the number of electrons passing a unit area parallel to the phantom surface. As the fluence ϕ and planar fluence ϕ_p are related to each other by:

$$\phi = \frac{\phi_p}{\cos \vartheta}, \quad (3.4.4)$$

where ϑ is the angle between the electron trajectory and the normal of the phantom surface, an increase of the average value of ϑ is accompanied by an increase of ϕ and hence in dose over the first few centimetres of the depth dose curve where ϕ_p is essentially constant as absorption of electrons is negligible in this region.

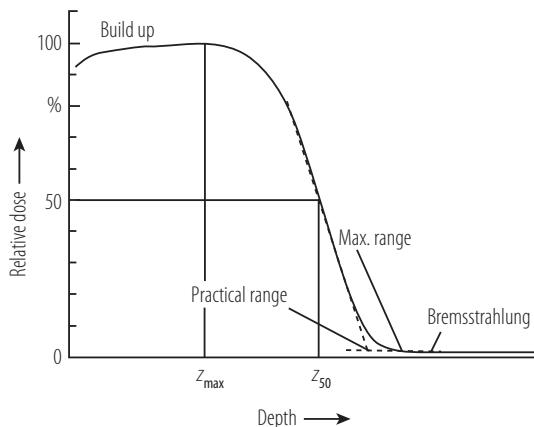


Fig. 3.4.2 Schematic representation of a depth dose curve for high-energy electron radiation. Important beam parameters are shown (see text).

The following parameters can be derived from a depth dose curve of an electron beam. The depth at which the dose maximum is located is termed z_{\max} , the point at which the dose amounts to 50 % of its value in the maximum is called R_{50} and the intersection of the linear extension of the descending part of the depth dose curve with the linear extrapolation of the so-called bremsstrahlung background (see Fig. 3.4.2) is called practical range R_p . At depth R_p essentially all electrons have lost their energy completely and are at rest. The non-zero value of the dose for depths greater than R_p is produced by contaminant bremsstrahlung which originates from the accelerator and also from the water phantom itself. The quantity R_{50} can be used to obtain a good estimate of the average electron energy in the entrance plane of the phantom \bar{E}_0 according to [65Har]:

$$\bar{E}_0 = 2.33 R_{50}. \quad (3.4.5)$$

The average energy at depth z inside the phantom, $\bar{E}(z)$ can be calculated by:

$$\bar{E}(z) = \bar{E}_0 \left(1 - \frac{z}{R_p} \right). \quad (3.4.6)$$

It is important to realise that the depth R_{50} is the depth at which the absorbed dose to water attains 50% of its maximum value, where the dose must not be equated to the indication of an ionization chamber. As the average energy of the electrons continually diminishes with increasing depth, the water-to-air stopping power ratio takes different values at different depth. According to the Bragg-Gray theory (see Sect. 3.2.2.3) the stopping power ratio relates the absorbed dose to water to the dose in the air of the cavity. To a good approximation the following numeric equations take account of this effect [95Din]:

$$R_{50} = 1.029 R_{50,\text{ion}} - 0.06 \text{ cm}, \quad R_{50,\text{ion}} \leq 10 \text{ cm}, \quad (3.4.7)$$

$$R_{50} = 1.059 R_{50,\text{ion}} - 0.37 \text{ cm}, \quad R_{50,\text{ion}} > 10 \text{ cm}, \quad (3.4.8)$$

where $R_{50,\text{ion}}$ is the depth in cm in which the ionization chamber has an indication amounting to half of that in the dose maximum.

3.4.2.3 Hadron beams

The diversification in range and direction of individual particles through scattering processes is denoted by the term straggling. Depending on the feature considered a distinction is made between range and directional straggling. The extent of straggling has a pronounced effect on the shape of the depth dose curve. In the beams of protons and also heavier ions, which are characterized by a weak straggling, three effects come together: (a) each individual particle loses almost identical amounts of energy per distance traveled, (b) this energy loss is accompanied by an increase in stopping power and (c) for a pencil beam all particles 'stay together' close to the beam axis. The result of these three effects is a sharp increase in the absorbed dose immediately before the very end of the range of the particles. This increase is termed Bragg-peak, an example of which is shown further down in Fig. 3.4.6 in Sect. 3.4.4.4.2 for the case of protons with a primary energy of 235 MeV.

3.4.2.4 Brachytherapy

For the dosimetry in brachytherapy the characterization of the source strength is of particular importance. Also in brachytherapy, the absorbed dose to water is the dosimetric quantity of interest. However, techniques for a direct measurement of the absorbed dose to water in close proximity to the source, e.g. at 1 cm, are currently under development by several metrology institutes, see for example [09Bam]. Until the results of these developments will be routinely available to the clinical user the source strength of photon emitting sources is characterised in terms of the air kerma (rate). Two quantities of this kind are being employed: the air kerma strength and the reference air kerma rate. The latter is the air kerma rate at a distance of 1 m from the source in a direction perpendicular to the source axis. In this definition the

contribution to the air kerma rate from scattered radiation produced in the air or by the walls of the measurement room is not taken into account, the term air kerma rate *in vacuo* could be used. The air kerma strength is reference air kerma rate multiplied with the ratio $(d/d_0)^2$, where d is the distance from the source and d_0 is 1 m. As the source can be practically considered as point source and in the absence of scattering the air kerma rate decreases quadratically with increasing distance, consequently the product

$$\dot{K}_a(d) \left(\frac{d}{d_0} \right)^2$$

is constant.

The source strength of β -radiation emitting sources is characterized in terms of absorbed dose to water. The reference value of the absorbed dose to water is usually determined in a water-equivalent phantom of a solid material.

3.4.3 Dosimetric equipment for radiation therapy

H.-M. KRAMER

A measurement of the 'amount' of radiation can, at least in principle, be based on whatever effect radiation has on matter. However, not all possible effects correlate sufficiently well with biological effects, a fact which excludes the use of many of them. Today, it seems widely agreed that the absorbed dose to water is well correlated with biological effects. The absorbed dose is defined as the amount of energy deposited in a volume element divided by the mass of this volume element. If the volume element consists of water, we speak of the absorbed dose to water (see Sect. 3.2.2.3). Water is chosen as the reference material because its radiation transport properties resemble quite closely those of many tissues and organs of the human body. A measurement of the absorbed dose to water according to its definition is usually performed only at metrology institutes. A device for measuring a physical quantity according to the definition of this quantity is called a primary standard. It is sometimes referred to as an absolute measurement, although this term may be misleading.

In metrology institutes primary standards for the absorbed dose to water have been developed on the basis of calorimetry and ionometry. In the first category of instruments the temperature rise induced by the ionizing radiation is measured either in graphite [04Pal, 02Nut] or in water [06Kra], while the second category is based on the production of charge carriers of opposite sign through ionization processes in ionization chambers [93Bou]. As the treatment of primary standards goes beyond the scope this contribution only a very few words will be devoted to this subject for the example of a water calorimeter.

At a given point in a water phantom, the absorbed dose to water is proportional to the temperature rise caused at this point by the radiation. The factor of proportionality is the specific heat capacity of water. Thus the temperature rise in water caused by an absorbed dose of 1 Gy = 1 J/kg is about 0.25 mK or, alternatively, a temperature rise of 1 mK requires a dose of about 4 Gy. From the above, two requirements for the operation of a water calorimeter are evident: small temperature increases need to be measured accurately and, as the dose is a quantity defined at a point, heat conduction phenomena need to be taken into account quantitatively.

The small temperature rise in water caused by typical therapeutic doses suggests immediately that a water calorimeter is not suited for routine measurements in practice. This calls for other, simpler methods of dosimetry. For photon and electron radiation and for energies up to a few tens of MeV the primary step in the transfer of energy from radiation to matter consists essentially of ionization and electronic excitation processes. It is therefore not surprising that the method the most frequently employed for a quantitative detection of ionizing radiation is based on the ionization of a gas inside an ionization chamber. In most cases the chamber volume is filled with air, which is (a) simple and has (b) the advantage that air has, apart from its density, radiation transport properties similar to those of soft tissue. The electric field established inside an ionization chamber by applying a polarizing voltage separates the charge carriers of opposite sign created by the radiation. The charge carriers migrate to their respective

electrode and an electrical charge is measured. Fig. 3.4.3 gives a schematic drawing of an ionization chamber and the electric circuit for current (left) and charge measurements. By multiplying the charge collected by the W -value, i.e. the average energy required for the production of an ion pair, (see Sect. 3.2.1.5) one obtains the amount of energy deposited in the gas volume of the cavity, which, to a first approximation, is proportional to the dose at the point of measurement, in absence of the chamber. For air and for electron energies above 5 keV the W -value is independent of the energy of the electron and amounts to $W = 33,97 \text{ eV}$. The energy independence of the W -value is of high practical importance as it allows dose measurements with ionometric methods without having knowledge of the energy of the ionizing particles.

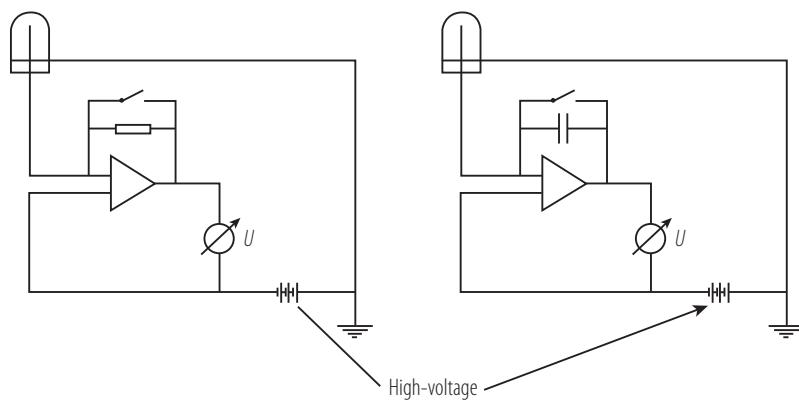


Fig. 3.4.3 Schematic drawing of the electrical circuit for measuring dose rate (left) and dose (right). Ionization chamber at the top left in each diagram.

Although an ionization chamber of known volume may be used for a measurement of the absorbed dose to water according to the definition of this quantity [93Bou], this is nowadays practically never done outside metrology institutes. Clinical dose measurements are performed according to international [00IAE, 01ICR] or national dosimetry protocols [96IPE2, 99AAP, 03IPE, 08DIN, 08NCS]. Invariably, these protocols call for the use of a reference ionization chamber which is appropriately calibrated in a reference field, which is usually ^{60}Co .

Reference ionization chambers are known in two different varieties: (a) as so-called thimble type chambers, usually cylindrical in shape with a collector electrode on the chamber axis also referred to as compact chambers and (b) parallel-plate chambers (see Fig. 3.4.4). Dose measurements in photon fields and in high energy electron fields are usually conducted with thimble type chambers. Measurements in electron fields, in particular those with mean energies below 10 MeV should be performed with parallel plate chambers. Reference chambers have typically a volume of around 0.5 cm^3 .



Fig. 3.4.4 Photography of ionization chambers. From top to bottom: thimble type chamber TM34001 with protective cover, 0.125 cm^3 , (PTW); thimble type chamber FC65-G with protective cover 0.65 cm^3 , (Iba); waterproof parallel-plate chamber TM 34001, 0.35 cm^3 (PTW).

This limits the spatial resolution of dose measurements, a feature which may be important for the determination of relative dose distributions inside a water phantom where steep dose gradients may occur,

especially for some forms of modern radiotherapy. For this kind of measurement one often uses also uncalibrated detectors of small volumes. By taking the reading of such a device at a point at which the dose has been determined each point in a relative dose distribution can be associated with a dose value, if needed.

In the following, a brief overview of detectors employed in a clinical environment is presented. Generally, a distinction is made between what is often termed active and passive devices. With the first term, a group of instruments is meant with a direct display of the momentary dose rate or the accumulated dose. Examples of such devices are ionization chambers or semiconducting detectors in connection with a suitable electrometer. The second term denotes a group of instruments where a probe accumulates the dose during the irradiation. After completion of the irradiation the value of the dose is obtained by means of a suitable read out device. In view of the dose information the read out process may either be destructive, e.g. in thermoluminescent dosimetry, TLD, or non-destructive (e.g. alanine, radio-photoluminescence RPL, radiochromic film). Passive dosimeters consists of the combination of detector, read-out device, possibly software and conditioning devices for the detector as for instance a heater for TLDs.

Small volume ionization chambers, also called microchambers, for relative dosimetry belong to the category of active devices employed for 3D dose mapping. They have a measuring volume between around 10 and 100 mm³ and can be used directly in water without a water protective sheath. Usually, they are not calibrated. It is convenient to position these devices inside the water phantom by means of a computer controlled 3D translational device, which allows a completely automated acquisition of 2D or 3D dose maps. Semiconductor detectors can be used in a similar set up. Their measuring volume is still considerably smaller. The thickness of the depletion layer may be as thin as a few micrometers and they have a cross-sectional area of typically up to a few square millimeters. A disadvantage of this kind of detector is that its response, i.e. the signal per unit dose changes appreciably with the accumulated dose. A further disadvantage is that the ratio of stopping powers - semiconductor material to water - may change appreciably with electron energy. Because of their better properties in view of this parameter diamond detectors, which have a similar spatial resolution, are sometimes preferred.

Two further kinds of active detectors should be mentioned in this context: liquid ionization chambers and scintillators. In the first kind of devices the sensitive volume is filled with a liquid, which in unirradiated form is insulating. By irradiating it charge carriers of opposite sign are generated and in the presence of an electrical field they migrate to the electrodes where they are collected and measured. The principle of liquid ionization chambers is much the same as that of gas filled ionization chambers with the difference that the density of the sensitive volume is larger by around three orders of magnitude and the *W*-value of the liquid(s) used is smaller than that of air. Both effects have the result that in a given radiation field the signal per unit volume from a liquid ionization chamber is much larger than that from a gas filled ionization chamber. This allows the construction of detectors which unite in themselves a high sensitivity and a high spatial resolution.

Certain plastic materials have the property of emitting scintillation light when exposed to ionizing radiation. By attaching a scintillator with a volume of typically 1 mm³ to a light guide which is connected to a photo-multiplier one obtains a dosimeter with a highly tissue-equivalent detector. A disadvantage of this type of dosimeters is the fact that Cherenkov radiation is generated in the light guide when it is hit by electrons with a speed above the speed of light in this material. In the literature two remedies for this drawback have been reported: By performing some kind of spectroscopic analysis the scintillation light can be distinguished from the Cerenkov light [05Fre]. Alternatively, it has been proposed to build a device with two essentially identical channels, with the difference that only one of them is actually furnished with a scintillator. By forming the difference signal between the two channels one obtains the signal just from the scintillator [00Bam].

A 2D dose profile, also in-phantom, can be obtained in one single exposure by means of a 2D detector array, which is a quadratic arrangement of 25x25 to 30x30 detectors that are either ionization chambers or solid state detectors. The advantage of the high speed of data acquisition of such devices is somewhat compensated by the relatively poor spatial resolution given by the inter-detector spacing of typically 1 cm. The spatial resolution may be improved by making a sequence of 4 exposures, with a translation of the detector after each exposure by half an inter-detector distance along the x and y directions and along the diagonal, a method which is still fairly quick.

A relative indirect method of dose mapping is based on portal imaging systems. These devices use the radiation transmitted through the patient/phantom for imaging by relating, for a given set up of field size and radiation quality, the brightness of a certain pixel or group of pixels to the dose at a point "upstream" in the patient/phantom.

Several systems of passive detectors find application in dose mapping. Some of them have imaging properties like silver halide and radiochromic film, storage screens, others are just probes for measuring the dose, sometimes *in vivo*, or at a given point inside a phantom.

Dose mapping by means of conventional silver halide films is a very quick method which provides a high spatial resolution. By sealing the film with a watertight foil, measurements can also be made in a water phantom. Attention should be paid to a homogeneous development of the film. One method of verifying whether the quality of the film development is acceptable is to expose a film of sufficient size to a homogeneous field. This can be achieved by placing the film free in air, preferably at a distance of 2 m or 3 m from the source. In order to make sure that no artifacts of inhomogeneity are caused by a possibly spatially varying fluence of secondary electrons the film is best placed immediately behind a build up layer which can, for instance, be realized by a polymethyl-metacrylate (PMMA) plate 5 mm in thickness or somewhat more.

The exposed and developed film is analyzed in terms of its optical density. The optical density is measured with a densitometer or, if a high resolution is required, with a microdensitometer. The optical density, d , is defined by $d = -\log_{10}(\phi/\phi_0)$, where ϕ_0 and ϕ are the incident and transmitted luminous fluxes. An example of the well known sigmoid shaped curve representing the relation between optical density and dose is given in Fig. 3.4.5. Prior to the conversion from optical density to dose such a curve needs to be determined for the prevailing conditions, i.e. radiation quality, kind and status of the film developer and the densitometer used.

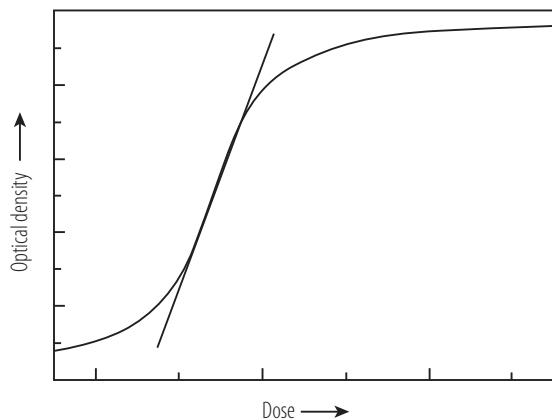


Fig. 3.4.5 Variation of optical density with increasing exposure. The slope of the tangent at the point of inflection is referred to as the sensitivity.

Another form of obtaining optical images of dose profiles is by means of radiochromic films, sometimes also referred to as Gaf-chromic film. These films consist of an active layer sandwiched between two plastic cover sheets, which undergoes a polymerisation when exposed to ionizing radiation. The incidence of the radiation causes directly an increase in optical density; no development is required. In a calibration process similar to that described above for conventional films, the combination of radiochromic film, scanner and radiation quality used needs to be calibrated in order to enable the conversion of optical density into dose. Again attention is required in order to avoid interpreting spatial response inhomogeneities as dose differences. This needs to be done for the scanner/film combination actually used. Radiochromic films are also sensitive to visible and ultraviolet light. They must be stored in darkness.

As radiological departments in clinics switch progressively away from silver halide films to digital imaging, storage screens are also used for dose mapping in radiotherapy. As typical image receptor dose values are one or two orders of magnitude smaller than the dose to the treatment volume in one session storage screens different from those used in diagnostic radiology are needed for therapeutic applications. In storage screens, the ionizing radiation creates a latent image which is read out by a laser scanner. The

incident laser light stimulates the emission of light of a certain wavelength, whose quantity is measured, digitized and attributed to the coordinates of the point of incidence of the laser light. In this way a digital image is generated. Unlike in the case of silver halide films the relationship between the latently stored light and the dose is linear over several orders of magnitude, which facilitates the conversion of signal to relative dose values.

Thermoluminescent dosimetry (TLD) and alanine dosimetry belong to the category of passive dosimeters. In the first type of detectors, electrons are trapped at metastable electronic interband levels which they reach through the irradiation. By heating the probe after the exposure, the electrons leave their metastable states making transitions to the ground state by emitting light which is recorded by means of a photomultiplier. The dose information of the probe is (almost completely) destroyed by the read out process. To regenerate the TLD-probes they are subjected to a well defined heating and annealing cycle. The TLD-probes, the read-out device, the algorithm with which the amount of light is converted to the value of the dose, the heating/annealing apparatus together with the heating/annealing procedure represent the TLD-system as a whole to which the calibration pertains. This calibration must not be used after change of any individual component of the system. For further information on the employment of TLD for clinical application see e.g. [92Fei].

In alanine, a simple protein molecule, free chemical radicals are generated by exposing it to ionizing radiation. Analyzing the irradiated probe by electron paramagnetic resonance a signal proportional to the number of radicals generated is obtained. The signal, divided by the mass of the probe, is proportional to the concentration of the free radicals, which, in turn is proportional to the absorbed dose in the alanine. The read-out process does not destroy the dose information. Similar to the case of TLDs described above, the ensemble of the probes, their manufacturing process, the EPR-spectrometer and the method of converting the EPR-signal to the value of the dose represent the alanine dosimeter. A once established calibration can only be used as long as no component is modified. For further information on alanine dosimetry see e.g. [96Sha, 05Ant, 06Ant]. A summary of detector properties is given in [Table 3.4.2](#).

Table 3.4.2. selected radiation detectors are listed together with their main field(s) of application.

Measurement device/technique	Long-term stability	Spatial resolution	Speed for obtaining 2D or 3D dose maps	Temperature dependence
Thimble type ionization chamber	++	-	-	++
Parallel-plate chamber	++	-	-	++
Micro chamber	+/-	+	-	+
Liquid ionization chamber	+/-	+	-	+/-
Semiconducting detectors	-	+	-	-
Scintillator	-	+	-	-
Thermoluminescent dosimetry	+/-	+/-	-	+
Alanine dosimetry	+	+/-	-	+/-
2D detector array	+/-	+/-	+	+
Silver halide film	-	++	++	+/-
Radiochromic film	-	++	++	+/-
Storage screen	-	++	++	+/-

3.4.4 Dosimetry for teletherapy

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3.4.4.1 Photons and electrons

H.-M. KRAMER

This Section is devoted to what is normally referred to as reference dosimetry, i.e. the determination of the absorbed dose to water under clinical reference conditions. The dose value obtained in this process plays a central role as an input parameter for the treatment planning. It is worth noting here that the clinical reference conditions are usually different from the reference conditions at a calibration laboratory. For further information on clinical dosimetry the reader is referred to [05IAE] and [07May].

The model equation according to which the absorbed dose to water D_W is determined in photon and electron beams is:

$$D_W = N \cdot (M - M_0) \cdot \prod k_i \quad (3.4.9)$$

where N is the calibration factor of the dosimeter, usually pertaining to ^{60}Co radiation, M and M_0 are the dosimeter readings (as taken from the display) in the presence and absence of the beam and k_i the correction factors. The correction factors k_i take into account that the conditions under which the dosimeter is used are usually different from the conditions under which it is calibrated, the so-called reference conditions. Each factor k_i takes care of the influence quantity i . Defining the response R of a dosimeter as the ratio of the dosimeter reading and the stimulus, i.e. the dose (rate), a correction factor is obtained by the equation $k = R_{\text{ref}}/R_{\text{user}}$ where R_{ref} and R_{user} are the dosimeter's response under reference and user conditions.

For clinical dosimetry the product of correction factors takes the following form:

$$\prod k_i = k_p \cdot k_p \cdot k_s \cdot k_h \cdot k_r \cdot k_T \cdot k_Q \cdot k_{\text{NR}}, \quad (3.4.10)$$

where the individual correction factors take care of

- k_p : density of air
- k_p : polarity effect of the ionization chamber
- k_s : incomplete charge collection
- k_h : humidity of the air
- k_r : displacement effect
- k_T : temperature effects other than the density of air
- k_Q : energy dependence of response and
- k_{NR} : deviations of the conditions of measurement from the clinical reference conditions

In the following paragraphs it will be assumed that the calibration factor pertains to the radiation of ^{60}Co . Should the calibration pertain to another radiation quality the respective values of correction factors for this radiation quality need to be taken.

3.4.4.1.1 Leakage current

Due to finite electrical resistance of the insulation in the ionization chamber, cable, connectors and in the electrometer itself and possibly through ionization processes caused by the environmental radiation an ionization chamber assembly has usually a non-zero reading in the absence of the beam. In order to make sure that the charge produced by these processes does not enter into the determination of the dose the additive correction M_0 is introduced in eq. (3.4.9). As the leakage current can be influenced by numerous effects, e.g. temperature, humidity, mechanical pressure on connecting cables, etc., the determination of M_0 should be an integral part of a dose determination.

3.4.4.1.2 Density of air

The calibration factor of a dosimeter refers to the density of air under reference conditions. For temperatures and pressures different from the reference conditions the mass of the air in the chamber cavity varies, while the dose applied to this gas remains unchanged. As the number of charge carriers produced in the measuring volume is proportional to its mass the chamber reading needs to be corrected in order to apply the calibration factor pertaining to the reference air density. Under atmospheric conditions at temperature T and pressure p the factor k_p is given by:

$$k_p = \frac{T \cdot p_0}{T_0 \cdot p}, \quad (3.4.11)$$

where T_0 and p_0 are the reference values for temperature and pressure. These reference values need to be taken from the calibration certificate. It should be observed that T and T_0 are absolute temperatures in Kelvin. While there is widespread agreement between calibration laboratories that $p_0 = 1013.25$ hPa, i.e. normal atmospheric pressure at sea level, there is some diversity in the values for T_0 which requires the user's attention.

3.4.4.1.3 Polarity effect

The polarity effect denotes the property of ionization chambers that the magnitude of the chamber reading changes when, under otherwise unchanged conditions, a polarizing voltage of opposite polarity is applied. As this effect depends on the constructional details of the chamber (materials, coatings etc.) it is corrected empirically by means of the following formula:

$$(k_p)_Q = \frac{\left(\frac{|M_1| - |M_2|}{|M_1|} \right)_Q}{\left(\frac{|M_1| - |M_2|}{|M_1|} \right)_{Co}} \quad (3.4.12)$$

where M_1 and M_2 are the chamber readings at the polarity usually used and at the opposite polarity, the magnitude of the polarising voltage being identical in both cases. All readings are normalised to the monitor reading to account for possible dose rate fluctuations during the different exposures. The subscripts Q and Co denote the radiation qualities employed by the user and for the chamber calibration. When the polarity effect of an ionization chamber is given in the calibration certificate this value can replace the expression in the denominator in eq. (3.4.12). The polarity effect depends on the radiation quality. When no information on the polarity effect for ^{60}Co -radiation is available and the user does not have this radiation quality at his disposal it is recommended to determine the denominator at the bremsstrahlung quality with the lowest energy provided by the linear accelerator, i.e. preferably 4 MV or possibly 6 MV.

3.4.4.1.4 Incomplete charge collection

To a certain extent the charge carriers separated from each other in the course of ionization processes recombine with each other before they reach their respective collecting electrodes of the ionization chamber. The result is that the magnitude of the charge actually measured is smaller than the charge produced. According to Boag [66Boa, 87Boa] one can distinguish between initial and general recombination.

Initial recombination denotes the process that charge carriers separated from each other by one single ionizing particle recombine with each other. Such events are also referred to as intra-track recombination. The magnitude of this process does not depend on the dose rate of the radiation. For loosely ionizing radiation like electrons the initial recombination can usually be neglected. For densely ionizing radiation like protons and in particular heavier ions the initial recombination may not be negligible.

In general recombination charge carriers of opposite sign recombine with each other on their way to their respective collecting electrode. The incomplete charge collection can either be taken into account by an empirical method based on so-called Jaffé-plots or alternatively by theoretical means mainly developed by Boag [66Boa, 87Boa].

The initial recombination is given by

$$\frac{1}{q} = \frac{1}{q_0} + \frac{\text{const}}{U} \quad (3.4.13)$$

where q and q_0 are the charges actually measured and produced and U is the bias voltage. In the case of pulsed radiation as it is produced by linear accelerators eq. (3.4.13) can also be applied for general recombination. The general recombination for continuous radiation is given by:

$$\frac{1}{q} = \frac{1}{q_0} + \frac{\text{const}}{U^2} \quad (3.4.14)$$

In Jaffé-plots $1/q$ is plotted vs. $1/U$ for initial recombination and for the total recombination for pulsed radiation. For general recombination of continuous radiation a plot of $1/q$ vs. $1/U^2$ is made. By extrapolating the plots to $1/U = 0$, i.e. to infinitely high bias voltages one obtains the desired quantity q_0 . When using Jaffé-plots one has to make sure that the extrapolation is made exclusively from a region where there is a linear relationship between $1/q$ and $1/U$ or $1/U^2$. Non-linear dependences can for instance be the result of gas amplification which may occur at high bias voltages in which case the charge carriers are sufficiently accelerated so as to create new charge carriers in collisions with the gas molecules.

According to Boag [87Boa] the collection efficiency, i.e. the inverse of k_s , is given in pulsed beams by

$$f(u) = \frac{1}{u} \ln(1+u) \quad (3.4.15)$$

where the variable u is defined as:

$$u = \mu \frac{q_0}{U} A_I \quad (3.4.16)$$

The coefficient μ depends only on the kind of ions and A_I is a double integral taking into account the path of the ions, which for homogeneously exposed parallel-plate chambers is equal to the square of the plate spacing. By definition, the charge measured, q , and produced, q_0 , are related to each other by $q_0 = q/f$. Inserting this into eq. (3.4.16) and making a variable transformation $v = f \cdot u = \ln(1+u)$ one obtains:

$$f = \frac{v}{e^v - 1}, \quad (3.4.17)$$

an equation relating the charge measured

$$q = \frac{v \cdot U}{\mu \cdot A_I}$$

and the collection efficiency f to each other. Usually approximations to eq. (3.4.17) are used for practical purposes by making a Taylor expansion. One such form applies to parallel-plate chambers, for which the saturation correction takes the form of the numeric equation:

$$k_s = 1 + 0.54D \frac{s^2}{U} \quad (3.4.18)$$

where D is the average dose per pulse in mGy, s the plate separation in mm and U the polarizing voltage in V.

The corresponding numeric equation for continuous radiation is:

$$k_s = 1 + 2.4 \dot{D} \frac{s^4}{U^2}, \quad (3.4.19)$$

where \dot{D} is the dose rate in units of Gy/s, s the plate separation in mm and U the polarizing voltage in V.

3.4.4.1.5 Humidity of air

In total, the factor k_h covers three different effects. In view of radiation transport, air saturated with water at approximately room temperature differs in three ways from dry air at the same temperature: its density is lower by about 1 %, the W -value (see Sect. 3.2.1.5) is lower by about 0.9 % and the mass-collision stopping power is about 0.2 % larger. Over the range of humidity normally encountered in the environment of a linear accelerators, which is typically in the range from 40% to 70% relative humidity, the three effects compensate each other to a high degree so that under these conditions no correction needs to be employed, provided that the calibration factor given in the calibration certificate refers to a relative humidity of (around) 50%. For more detailed information concerning the three physical effects the reader is referred to reference [88Rog].

3.4.4.1.6 Radiation quality

Photon radiation

In Sect. 3.4.2.1 the radiation quality index Q was defined as the tissue phantom-ratio $\text{TPR}_{20,10}$ (eqs. (3.4.2) and (3.4.4)). Once the index Q is determined experimentally it may be used in order to take care of the energy dependence of response of the detector used. The following paragraphs describe the procedures to be followed. As modern protocols for photon dosimetry provide the required information for thimble type chambers only, this restriction will also be adhered to in what follows. For reasons of completeness it should be mentioned, however, that experimental evidence is coming up indicating that reference dosimetry in megavoltage X-ray beams may also be carried out by means of parallel-plate chambers [10Kap].

It is convenient to split the correction factor for the energy dependence of the response, k_Q , into two factors, i.e.:

$$k_Q = k'_Q \cdot k''_Q, \quad (3.4.20)$$

where the first one takes into account the properties of the radiation field in terms of the ratio water to air of the stopping powers according to:

$$k'_Q = \frac{(s_{w,a})_Q}{(s_{w,a})_{Co}} = \frac{\left(\frac{s_w}{s_a}\right)_Q}{\left(\frac{s_w}{s_a}\right)_{Co}}. \quad (3.4.21)$$

The subscripts Q and Co refer to the user's radiation quality and to the one used for the calibration of the chamber. Values of k'_Q can be taken from Table 3.4.3.

Table 3.4.3. Values of $k'_Q = (s_{w,a})_Q/(s_{w,a})_{Co}$ as a function of the radiation quality index Q for a depth of 10 cm and a field size of 10 cm x 10 cm at the depth of measurement. Values taken from [94And] and divided by the mass-stopping power ratio 1.133 for ^{60}Co gamma radiation.

Q	0.50	0.53	0.56	0.59	0.62	0.65	0.68	0.70	0.72	0.74	0.76	0.78	0.80	0.82	0.84
k'_Q	1.002	1.001	0.999	0.997	0.996	0.992	0.988	0.983	0.979	0.974	0.967	0.960	0.952	0.943	0.932

The factor k''_Q takes into account the deviations of a real ionization chamber from a perfect Bragg-Gray cavity. This is expressed by means of the so-called perturbation factors according to:

$$k''_Q = \frac{p_Q}{p_{Co}} = \frac{(p_{wall} p_{dis} p_{cel})_Q}{(p_{wall} p_{dis} p_{cel})_{Co}}, \quad (3.4.22)$$

where indices indicate the perturbations caused by the chamber wall, the so-called displacement effect and the centre electrode. By definition k''_Q is one for a perfect Bragg-Gray cavity. The essence of

[eq. \(3.4.22\)](#) is the following: the calibration factor provided in the calibration certificate can be interpreted as the product of the calibration factor, N^{BG} , of a perfect Bragg-Gray cavity with a volume identical to that of the chamber actually used and the factor p_{Co} . Thus dividing the calibration factor provided in the calibration certificate by p_{Co} one obtains N^{BG} . If now N^{BG} is multiplied by p_Q and the stopping power ratio according to [eq. \(3.4.21\)](#) one obtains what could be termed the calibration factor for the user's radiation quality Q .

A non water-tight ionization chamber can only be brought into a water phantom by means of a protective sheath. In the general case, where the chamber wall and the sheath are made of different materials, it has been suggested that the factor p_{wall} can be calculated according to [77Alm, 85Gil, 85Han]:

$$p_{wall} = \frac{\alpha \cdot s_{wall,a} \cdot t_{W,wall} + \tau \cdot s_{sheath,a} \cdot t_{W,sheath} + (1 - \alpha - \tau) \cdot s_{W,a}}{s_{W,a}}, \quad (3.4.23)$$

where t is the ratio of mass-energy absorption coefficients, the subscripts are as above or self-explaining and the parameters α and τ denote the relative fraction of the total ionization in the cavity caused by electrons generated in the chamber wall and in the sheath. For the special case of a water proof chamber where no sheath is used the parameter τ is set zero. Recent Monte Carlo calculations suggest that the model associated with [eq. \(3.4.23\)](#) may be an oversimplification [10Wul].

The displacement perturbation factor p_{dis} will be dealt with in the following Section. The determination of the factor p_{cel} requires a simulation of the ionization chamber with and without the central electrode by means of Monte Carlo methods, see e.g. [93Ma].

Electron radiation

With one exception, modern dosimetry protocols on electron dosimetry require either the use of a thimble type chamber calibrated - usually - at ^{60}Co radiation quality or, alternatively, of a parallel-plate chamber that is metrologically linked to a thimble type chamber through the process of cross-calibration conducted by the user. An exception in the sense that electron dosimetry with parallel-plate chambers may also be based on a ^{60}Co calibration is given DIN 6800-2 [08DIN]. In the following a description of the dosimetry using a thimble type chamber will be given first followed by some guidance how to perform a cross-calibration. This Section will be concluded with some remarks on the use parallel-plate chambers possessing a ^{60}Co calibration factor.

As has been done for photon radiation it is also convenient for electron radiation to split the factor k_Q into the product of two factors according to [eq. \(3.4.20\)](#) (see above in this Section). As outlined in [Sect. 3.4.2.2](#) the user's radiation quality for electrons is specified in terms of R_{50} , i.e. the depth where the dose acquires 50% of its value in the dose maximum, which leads to:

$$k'_Q = \frac{(s_{w,a})_{R_{50}}}{(s_{w,a})_{Co}}. \quad (3.4.24)$$

Again, replacing in [eq. \(3.4.22\)](#) Q by R_{50} one obtains for electron radiation:

$$k''_Q = \frac{p_{R_{50}}}{p_{Co}} = \frac{(p_{wall} p_{dis} p_{cel})_{R_{50}}}{(p_{wall} p_{dis} p_{cel})_{Co}}. \quad (3.4.25)$$

For both thimble type and parallel-plate chambers it has been assumed that the perturbation factor for the chamber wall is one for electron irradiation [00IAE]. Newer studies on parallel-plate chambers suggest that p_{wall} may deviate from the value of one by as much as 1 %. For further information on this subject the reader is referred to the work by Zink and co-authors [11Zin] and references given therein. As has been done for photon irradiation the factor p_{dis} will be dealt with in the following Section. Information on p_{cel} is relative sparse, in ref. [00IAE] a value of $p_{cel}=0.998$ is given for a centre electrode of aluminium. For a number of chamber types table 37 in reference [00IAE] gives values of $(p_{cel})_{Co}$.

When using a parallel-plate chamber for which no value for k''_Q is available a process called cross-calibration [97IAE] needs to be performed by the user in order to obtain the missing information. To this end a dose measurement according to [eq. \(3.4.9\)](#) is performed with a thimble type chamber. The radiation

quality should be the highest electron energy available and a field size of at least 20 cm x 20 cm should be used. In the next step the thimble type chamber is replaced by the parallel-plate chamber that is exposed under identical conditions to the same dose as the thimble type chamber. This allows the determination of $N^{pp}(k_Q'')^{pp}$ by equating the doses

$$D_W^c = N^c \cdot (M - M_0)^c \cdot \left(\prod k_i \right)^c$$

and

$$D_W^{pp} = N^{pp} \cdot (M - M_0)^{pp} \cdot \left(\prod k_i \right)^{pp},$$

where the superscripts refer to the compact and the parallel-plate chamber. Then $N^{pp}(k_Q'')^{pp}$ is obtained according to:

$$N^{pp}(k_Q'')^{pp} = \frac{(k_\rho \cdot k_s \cdot k_p \cdot k_T)^c N^c M_Q^c}{(k_\rho \cdot k_s \cdot k_p \cdot k_T)^{pp} M_Q^{pp}} (k_Q'')^c, \quad (3.4.26)$$

which in turn allows the use of the parallel-plate chamber for measurements according to [eq. \(3.4.9\)](#).

3.4.4.1.7 Displacement effect

For this Section the terms point of measurement and reference point of a detector are of importance. The point of measurement is the point in the undisturbed phantom at which the absorbed dose is to be measured. The reference point of a detector is a point specified by the manufacturer of the device with respect to which the detector is located inside the phantom. Manufacturers of thimble type chambers usually define a point on the cavity axis as the reference point while for parallel-plate chambers they usually chose the centre of the entrance wall - cavity interface.

In an air-filled cavity located within an irradiated phantom the radiation is less strongly absorbed than in the surrounding phantom. This effect is essentially produced by the density differences between the air in the cavity and the phantom material. Because of the small density of the air it may be assumed, to a good approximation, that the absorbed dose is constant over the volume of the air cavity, while the absorbed dose inside the phantom follows the depth dose curve (see [Sects. 3.4.2.1](#) and [3.4.2.2](#) for photon and electron beams). As a consequence of the different dose distributions in the cavity of the chamber and in the phantom the properties of a detector with finite volume deviate from that of an ideal Bragg-Gray cavity as they were assumed in [eq. \(3.2.11\)](#) in [Sect. 3.2.2.3](#).

Consider a thimble type chamber placed with its reference point (on the cavity axis) at the point of measurement inside the phantom. Using [eq. \(3.2.11\)](#) one attributes the average dose absorbed in the cavity, multiplied by the ratio - phantom material to air - of the averaged stopping powers to the absorbed dose to the phantom material at the point inside the phantom where reference point of the chamber is located. Because of the negligible dose gradient inside the cavity and the negative gradient in the falling part of the depth dose curve in the phantom [eq. \(3.2.11\)](#) attributes too high a value of the dose to the point where cavity centre is positioned.

Two methods can be employed to overcome the problem: (a) a perturbation factor p_{dis} may be applied taking into account that the average absorbed dose in the cavity results in an overestimation of the absorbed dose in the (undisturbed) phantom at the point of measurement or (b) defining, in the coordinate system of the phantom, an effective point of measurement in such a way that [eq. \(3.2.11\)](#) may be employed without the factor p_{dis} . Once distance and direction between the reference point of the chamber and the effective point of measurement are determined it is convenient to define the effective point of measurement in the coordinate system of the chamber with respect to its reference point. This allows to locate the effective point of measurement of the chamber to point of measurement in the phantom and no further correction is needed.

How can one find the effective point of measurement? To do so, let us consider the example of the parallel-plate chamber whose walls consists of the phantom material. Assuming that scattering from the

chamber walls is negligible the differences in the depth dose curves in the undisturbed phantom and in the cavity increase from zero at the entrance window - cavity interface to larger values at greater depths in the cavity. Thus, the effective point of measurement of our model parallel-plate chamber is the centre of the entrance window - cavity interface.

For thimble type chambers the above considerations apply in principal as well. Because of the curved chamber wall - cavity interface they are however less straight forward. For thimble type chambers values for the separation of the effective point of measurement from the chamber axis between $0.4 r$ and $0.75 r$ are found [77Joh, 06Kaw, 08DIN 08NCS], where r is the cavity radius. The effective point of measurements is located 'upstream' relative to the chamber axis.

Locating the effective point of measurement (of the chamber) at the point of measurement (in the phantom) eq. (3.2.11) can be applied without using the factor k_r . In other words, the reference point of a thimble type chamber (assumed to be on the cavity axis) needs to be placed downstream by a certain fraction of r in order to use eq. (3.2.11) correctly. This method of detector positioning is very practical as it allows the measurement of a complete depth-dose curve starting from the build-up region down to large depths without the need for calculating any value of the factor k_r . For further studies the reader is referred to references [08Zin, 09Tes, 09Zin, 10Wan, 10Kaw].

In the discussion up to now it has been assumed for the argument's sake that one resorts to eq. (3.2.11). However, for clinical dose measurements this is, at least in general, not the case. Instead the model eq. (3.4.9) together with eq. (3.4.10) is used which is based on the use of a traceably calibrated ionization chamber. In the following some explanations are given on how to handle the displacement effect/the factor k_r when using detectors calibrated at a calibration laboratory. Furthermore, some essential differences are addressed in how the displacement effect is dealt with in some dosimetry protocols [99AAP, 00IAE, 08DIN].

If not all, then at least the overwhelming majority of calibration laboratories locate for the calibration of an ionization chamber its reference point at the point of measurement. The reason for doing so lies in the fact that using the effective point of measurement has some metrological weaknesses that have to do with model dependent ambiguities where the effective point of measurement is actually located relative to the reference point of the chamber.

For water as the phantom material and taking into account the displacement effect eq. (3.2.11) is modified to read:

$$D_W = D_a \frac{(\bar{s}/\rho)_W}{(\bar{s}/\rho)_a} p_{\text{dis}}. \quad (3.4.27)$$

Inserting D_a from eq. (3.2.12) into eq. (3.4.27) and postulating that D_W from eq. (3.4.27) and eq. (3.4.9) be identical one obtains the expression for the calibration factor of an ionization chamber with water-equivalent walls and with volume V as:

$$N = \frac{W}{e} \frac{1}{\rho V} \frac{(\bar{s}/\rho)_W}{(\bar{s}/\rho)_a} p_{\text{dis}}. \quad (3.4.28)$$

It has been assumed that the correction factors apart from k_r are all one and that $Q = (M-M_0)$. Assuming that the effective point of measurement of a thimble type ionization chamber is located 'upstream' relative to the reference point by half its inner radius, r , the factor p_{dis} is given by:

$$p_{\text{dis}} = 1 + |\delta| \frac{r}{2}. \quad (3.4.29)$$

where δ is the gradient of the depth dose curve at the point of measurement in the calibration field.

Eq. (3.4.28) shows that the calibration factor provided by a calibration laboratory contains the factor p_{dis} applicable to the gradient in the calibration field. The calibration factor pertaining to an ionization chamber located with its effective point of measurement can be derived from the calibration factor provided by the calibration laboratory by dividing the latter by p_{dis} .

Two principal options for dose measurements exist: (a) the detector's effective point of measurement is brought to the point of measurement, in which case D_W is obtained by:

$$D_W = N \cdot (M - M_0) \cdot \prod_{i \neq r} k_i / p_{\text{dis}} . \quad (3.4.30)$$

Using p_{dis} of the calibration field eq. (3.4.30) applies to measurements in any radiation field, regardless of the magnitude and the sign of the gradient of the depth dose curve is. In option (b) the detector is placed with its reference point at the point of measurement. In this case the factor k_r needs to be applied which is given by:

$$k_r = \frac{1 + |\delta_Q| r/2}{1 + |\delta_{Co}| r/2} . \quad (3.4.31)$$

For both options the dose gradient of the calibration field is needed. Therefore some calibration laboratories specify its value in the calibration certificate. If this information is missing, the dose gradient of another ^{60}Co -field should be used.

In dosimetry protocols [99AAP, 00IAE, 08DIN] both prescriptions on the positioning of an ionization chamber are found, where the kind of radiation and of the type of chamber play an important role. For any dose measurements according to a dosimetry protocol it is therefore essential to follow the specified positioning instructions meticulously. Only then one may use the numerical data of the correction factors provided in the protocol. On no account should one use a combination of numerical data of correction factors taken from different protocols.

So far the considerations apply primarily to photon radiation. Also for electron radiation the displacement perturbation needs to be taken into account. For thimble type chambers in electron fields p_{dis} is caused by electrons scattered from the chamber walls into the cavity, an effect that depends on the mean electron energy and the cavity radius. These effects have been studied by Johansson et. al. [78Joh]. The following function [08DIN] has been fitted to the data of Johansson [78Joh]:

$$p_{\text{dis}}(r, \bar{E}(z)) = 1 - 0.02155 \cdot r \cdot e^{-0.1224 \bar{E}(z)} . \quad (3.4.32)$$

Making use of the empirical relation [08DIN]

$$R_p = 1.271 \cdot R_{50} - 0.23 \quad (3.4.33)$$

and using eqs. (3.4.6) and (3.4.7) one can express the factor p_{dis} explicitly as a function of r , R_{50} and z according to:

$$p_{\text{dis}}(r, R_{50}, z) = 1 - 0.02155 \cdot r \cdot e^{-0.2852 \cdot R_{50} \left(1 - \frac{z}{1.271 \cdot R_{50} - 0.23} \right)} . \quad (3.4.34)$$

This Section should be concluded with a word of caution. For the treatment of the displacement effect a deliberate choice of simple models was made here in order to be able to explain the physical principles. From several recent studies by means of Monte Carlo evidence is coming up (see, e.g., [08Zin, 09Tes, 09Zin, 10Wan, 10Kaw]) indicating that the physics in real ionization chambers may be more complex than the treatment with analytical formula suggests.

3.4.4.1.8 Temperature effects other than the density of air

This correction factor covers any temperature effect that may have an influence on the response of an ionization chamber apart from that of the density of air, which is dealt with in Sect. 3.4.4.1.2. An example of such an effect could be the thermal expansion of some components of an ionization chamber that results in a change of the chamber volume. When an ionization chamber exhibits this kind of effects an attempt can be made to correct for these effects in an empirical way. In any case attention is required in order to be sure that the magnitude of the effect does not change in course of time. If this should be the case, one should consider replacing the unstable device by a detector not exhibiting such effects.

3.4.4.2 Non-reference conditions

H.-M. KRAMER

Dosimetry protocols provide information on correction factors and their numerical values for reference conditions. Some of the protocols, e.g. [00IAE, 08DIN] touch the issue of dosimetry under non-reference conditions, but they provide hardly any or no numerical values for correction factors applicable to such conditions. In this context non-reference conditions are to denote situations where measurements are performed at depths and/or field sizes different from their reference values and off-axis. This includes measurements under so-called Intensity Modulated Radiation Therapy (IMRT) conditions. For further information on IMRT see the review article by Bortfeld [06Bor]. In the following the term IMRT is used as a generic term which is to include also special forms of radiation treatment like TomoTherapy® (see review by Mackie [06Mac]) and with special radiation treatment machines like CyberKnife® and Lecksell Gamma Knife™.

In measurements under conditions different from reference conditions the properties of the radiation field in terms the spectral and angular distribution of particle fluences are different from what they are under reference conditions. These differences have an impact on the response of detectors.

One of the problems of providing guidance on measurements under non-reference conditions is the enormous variability of these conditions which makes it difficult to establish a finite number of representative situations each of which is to stand for a whole family of situations which all could be handled with one set of data. An example of a framework on how to handle dosimetry under IMRT conditions can be found in the article by Alfonso et. al. [08Alf].

Some photon radiation treatment machines like the CyberKnife® and the TomoTherapy® unit do not provide the possibility of establishing the conditions for determining the radiation quality index Q (see eqs. (3.4.2) and (3.4.4)) because these machines do not allow the realization of a 10 cm by 10 cm field required for the determination of Q . Even if they did, the Q -value obtained with these machines would not be applicable for a determination of the correction factor k_Q appropriate to this kind of machines, the reason being that these machines do not have a so-called flattening filter. In conventional linear accelerators such a device is located in the beam immediately behind the target and is shaped in such a way that field sizes up to 30 cm by 30 cm may be realized with a homogeneous dose rate over the entire cross-sectional area. The k_Q factors given in dosimetry protocols pertain to beams with flattening filters.

Attempts have been made to determine the radiation quality index (for a 10 cm x 10 cm field, see Sect. 3.4.2.1) for machines not capable of providing this field size. This is done by using an accelerator which allows a continuous variation of the field size under otherwise unchanged conditions. By determining a generalized radiation quality index $Q^g(l)$ as a function of the size of a square field of side length l a relation between $Q^g(l)$ and $Q^g(l = 10 \text{ cm}) \equiv Q$ is obtained which can be used to convert a value of $Q^g(l)$ for a certain value of l to Q [09Sau]. This approach for determining k_Q -factors for IMRT conditions appears to be justified as scientific evidence is coming up that - at least for field sizes between 3 cm and 10 cm side length - the response of frequently used ionization chambers is independent of field size [07Kra, 10Kra]. However one should be aware that for field sizes below a certain limit the k_Q -factors taken from dosimetry protocols [00IAE, 08DIN] on the basis of such extrapolated Q -values will lead to incorrect results because of the absence of lateral electronic equilibrium even on the beam axis with the consequence that stopping power ratio will take different - field size dependent values.

3.4.4.3 Kilovolt X-ray beams

H.-M. KRAMER

For some medical indications like, e.g., skin treatment, irradiation of joints, kilovolt X-ray beams are employed. Dosimetry of kilovolt X-ray beams follows in principle the formalism presented in Sect. 3.4.4.1. As the treatment volume is either the skin itself or not deeply seated, a depth of 2 cm or 3 cm is sometimes considered as the reference depth. A reasonably up to date code of practice for the dosimetry of kilovolt X-ray beams is the one prepared by the Institution of Physics and Engineering in Medicine and Biology in the UK [96IPE1]. It is based on the use of ionization chambers calibrated in

terms of the quantity air kerma and provides, for three separately treated energy ranges, the information on how to arrive at the absorbed dose to water as the quantity of interest. Some metrology laboratories are currently making efforts to provide in phantom calibrations in terms of absorbed dose to water for kilovolt X-ray beams [11Kra] which in turn could lead to the development of a native absorbed dose to water dosimetry code of practice.

3.4.4.4 Protons and heavier ions

S. VATNITSKY

3.4.4.4.1 Introduction

The advantages offered by proton and heavier ion beams compared to other radiation used in radiotherapy are based on three basic essentials. First, the low entrance dose and almost zero dose distal to the target results in the dose delivered to normal tissues relative to the dose delivered to target tissues being much lower than for high-energy photon and electron beams. Second, the lateral and distal dose gradients are higher enabling better protection of normal tissues. The third, which only applies to ions heavier than protons, is that a differential relative biological effectiveness with depth results in a therapeutically relevant dose in target tissues compared to surrounding normal tissues. Fig. 3.4.6, left shows a depth dose distribution for a 235 MeV monoenergetic proton beam. The dose is relatively constant with depth for most of its penetration, rises rapidly just before the particles lose all their energy, and then drops sharply almost to zero. Protons lose kinetic energy primarily in small discrete amounts as they undergo a multitude of interactions such as atomic ionization and elastic nuclear collisions and an initially monoenergetic beam of protons will thus acquire a small energy spread after passing through an absorber. Through this process the secondary electrons and delta rays are produced at all depths but the range of these electrons is quite small. This effect is important for ionization chamber dosimetry protocols where the effect of the secondary electrons produced outside the chamber cavity is neglected. Beyond the buildup region the most apparent process is the increasing stopping power. The other process is a decrease in number of protons due to nuclear interactions. Once the protons slow down and reach their peak energy deposition rate, the protons run out of energy and stop. However, not all protons stop at the same depth and so the depth dose curve just beyond the peak is not a vertical line but instead has a steep slope. The physical characterization of ions heavier than protons is similar in terms of dosimetry to that used with protons. However, the nuclear interactions products are in wider range and the ions themselves may undergo fragmentation as a result of the interaction with the target nuclei in addition to delta ray production. The important outcome is that the light projectile fragments have a larger penetration depth than the primary particles leading to a larger fragmentation tail beyond the Bragg peak compared to protons.

3.4.4.4.2 Beam delivery techniques

The beam that is delivered from the accelerator to the treatment room is monoenergetic, has a cross-section of several millimeters, and in order to be used clinically this beam should be spread out laterally and modulated in energy to make a uniform dose distribution to cover the Planning Target Volume (PTV). In a passive beam delivery method different thicknesses of material are inserted into the beam path to shift the range of protons in the patient to different depths and depth dose distributions are conformed to the distal surface of the PTV by the use of patient specific compensators. Alternatively, in active beam delivery methods the beams are extracted directly from the accelerator at the required energies and stacked at predefined layers using different scanning technologies. The resulting uniform high dose region (see Fig. 3.4.6 – right) is called a spread-out Bragg peak (SOBP). In addition, the lateral spread is needed to achieve uniform fluence in the beam cross-section to cover the PTV and this can be done by using scattering foils and patient specific apertures. An alternative way is to scan a pencil beam across the desired field using a non-modulated beam flux to cover the field with a uniform fluence or to

use modulated fluences that are delivered to different areas within the treatment field. Linear energy transfer (see Sect. 3.2.2.3) is higher with carbon ions than with protons resulting in a greater relative therapeutically relevant biological effectiveness (RBE). The RBE for carbon ions increases with depth therefore yielding a higher biologically relevant dose relative to the surface than with proton beams. This effect requires biological optimization in planning of carbon beam therapy when a biological dose distribution is used to characterize SOBP. It is important to realize that the therapeutically relevant RBE referred to in this Section is both conceptually and numerically different from that used in radiation protection (see Chapter 2 in LB Vol. VIII/4)

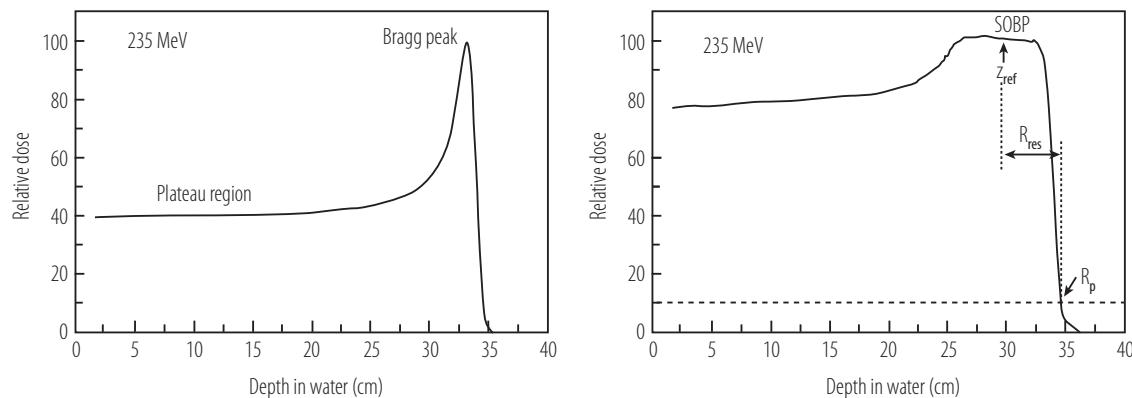


Fig. 3.4.6 Left: Percentage depth-dose distribution for a 235 MeV proton beam, illustrating the “plateau” region and the Bragg peak. Right: percentage depth-dose distribution for a modulated proton beam (indicated on the figure are the reference depth z_{ref} (middle of the SOBP), the residual range at z_{ref} used to specify the quality of the beam, R_{res} , and the practical range R_p) [00IAE].

3.4.4.4.3 Reference dosimetry methods

The lack of national and international dosimetry standards for protons and heavier ion beams historically made user operated calorimeters and Faraday cups (FC) the instruments of choice for the calibration of dose per monitor unit (MU) for these beams. In Report 78 the ICRU [08ICR] recognized that the major difficulties with both water and graphite calorimetry currently have largely been understood, thus recommended that, when available, calorimeters should be used as primary standards or, alternatively, to determine or to confirm the proton calibration factor of the ionization chambers. Due to the large but not sufficiently understood differences frequently observed between FC-dosimetry and calorimeters or ionization chamber dosimetry, a recommendation was made that FC should not be used as the sole method for calibration of clinical ion beams. Water is recommended as the reference medium for the determination of absorbed dose and for beam quality measurement.

Practically the dosimetry protocols implemented in proton and heavier ion beam therapy [98ICR, 08ICR, 00IAE] are based on ionization chamber dosimetry. Similar to high-energy photon and electron beam dosimetry these protocols include several major steps. First, the basic physics data to convert charge to absorbed dose for well characterized conditions are provided; second, the user is guided on how to correct the response of the ionization chamber for different influence quantities; third, the user is guided on how to obtain appropriate beam quality correction factors. Report 78 of the ICRU [08ICR] deals with protons only, but TRS 398 [00IAE] recommendations cover protons and heavier ion beams as well.

The use of the basic physics data can be summarized as follows. The W -values for protons in air from Jones [06Jon] that were included in report 78 of the ICRU [08ICR] showed that only the values deduced from the comparison of all available calorimetry and ionometry data should be used for the determination of the (W_{air}/e) -value for the proton ionization chamber dosimetry. The value thus determined is $(34.2 \pm 0.14) \text{ J}\cdot\text{C}^{-1}$, which is consistent with the value recommended in TRS 398 [00IAE] and is almost identical to the value of $34.3 \text{ J}\cdot\text{C}^{-1}$ given in AAPM Report 20 [86AAP]. Note that the (W_{air}/e) -values

recommended in TRS-398 for proton and heavier ion beams correspond to dry air although most facilities make measurements under humid air conditions.

Both TRS-398 [00IAE] and ICRU 78 [08IVR] adopted the proton stopping power ratios calculated with the Monte Carlo code PETRA [97Med] where the basic proton stopping powers are taken from Report 49 of the ICRU [93ICR]. These calculations include secondary electron transport and nuclear interactions. According to TRS-398 [00IAE], the mass-stopping power ratios and also the (W_{air}/e) -values for ions heavier than protons are assumed to be independent of the beam residual range because of the current lack of experimental data. The contribution of fragmented nuclei to the mass-stopping power ratios and (W_{air}/e) -values are also assumed to be negligible. Due to the larger mass, the elastic scattering of heavy ions is significantly reduced as compared to protons and lateral scattering can be also neglected. Constant values of the mass-stopping power ratio and (W_{air}/e) -value are therefore adopted currently and recommended in TRS-398 for all light-ion beams. These values are 1.130 and $(34.50 \pm 0.52) \text{ J}\cdot\text{C}^{-1}$, respectively.

The major physical processes that lead to recombination of charge carriers of opposite sign in proton and heavier ion beams are similar to those that occur in ionization chambers placed in high-energy photon beam and described in Sect. 3.4.4.1.4. However, the treatment of the initial and general recombination should be different due to the features of the particle accelerators. To guide practical users TRS-398 [00IAE] considers only pulsed beams and recommends in all cases to use a two-voltage method. Recent publications indicate that the recombination corrections can be overestimated by up to 2%, if the recommendation from TRS-398 [00IAE], which is only valid for pulsed beams, is followed without further considerations [06Pal]. Some cyclotrons produce short pulses (e.g. $\tau \sim 400 \mu\text{s}$) of particles but at such a high repetition rate that the beam is considered to be continuous. Most synchrotrons are pulsed, but at such a low frequency and with such a long extraction spill length that the extracted beam is also considered to be continuous. Therefore recombination corrections for ionization chambers in cyclotron or synchrotron –based delivery systems that use uniform beam scanning with energy stacking or small, high dose rate scanned beams should depend on the relationship between beam pulse duration and the ion transit time in the ionization chamber. If the beam pulse duration is of the same order of magnitude as the ion transit time in the ionization chamber, then the conditions pertaining to continuous beam irradiation are met [08Lor]. In all other cases the two-voltage method can be applied. To verify the applied corrections the charge collection efficiency of ionization chambers for scanning proton systems and especially for heavier ion beams should be assessed by calibration against a dose rate independent device, such as a calorimeter or Faraday cup. The recommendations on polarity effect corrections given in Sect. 3.4.4.1.3 are valid for proton and heavier ion beams.

3.4.4.4 Non-reference conditions

Water is also employed for relative dosimetry as it is easy to use in computerized scanning phantoms. Solid phantoms are convenient for relative measurements but the water-equivalent thickness of the material should be determined for each ion species. Active and passive detectors for dosimetry under non-reference conditions or for relative dosimetry of proton and heavier ion beams must have the appropriate sensitivity, energy and dose rate independence, and spatial resolution for each clinical dosimetry task. The time structure of the beam must also be considered for clinical dosimetry measurements. Dynamic beam delivery measurements require to accumulate the time-varying signal of the detector placed at a fixed point in a phantom over the full time of a scan; therefore the systems with active multiple detectors in one-, two-, or three-dimensional arrays can save time. Additionally a scintillating screen viewed by a CCD system may be considered as a very useful tool to visualize the delivered dose in both static and dynamic beam deliveries. It is possible to measure the dose in two dimensions at different depths with a position resolution of a fraction of millimeter, and with a reproducibility of the data of about 1% [05Ped]. Another advantage of the screen-CCD camera system is the immediate availability of the digital data that also makes it very useful in the QA procedures of scanning beams.

From the wide selection of detectors dealt with in Report 78 of the ICRU, ionization chambers, silicon diodes, radiographic and radiochromic films, diamond detectors, gels, scintillators and TLDs only ionization chambers are considered in many cases as the standard instrument. LET effects have a

substantial influence on the response of solid state detectors, gels, films etc. in ion beams. This physical property has to be taken into account and the behavior of each passive detector should be verified against an appropriate reference ionization chamber. Although diodes have many advantages as active detectors for dose measurements, any practical use of diodes should be carefully checked by comparing results with other detectors - a parallel-plate ionization chamber can be used as a reference for depth dose measurements and a small volume ionization chamber - for profile measurements. Silicon diodes also exhibit an increase of response with temperature and a significant loss of response with accumulated dose. In practice the increase in response with temperature needs to be determined for each new diode and checked periodically with the simultaneous check of response with accumulated dose.

Diamond detectors have excellent spatial resolution similar to diodes but due to their better physical properties can be calibrated in terms of absorbed dose to water and may also be used for absolute absorbed dose determinations in ion beams. A pre-irradiation of the diamond detector to 5 Gy - 15 Gy is required to stabilize its response. This pre-irradiation should be repeated each time the bias is turned off and back on. Depth doses and lateral profiles obtained with diamond detectors generally are in a good agreement with the data from small volume ionization chambers. Nevertheless the application of diamond detectors for absorbed dose determinations in low-energy proton beams requires the determination of the dose rate and LET correction factors.

Similar to high-energy photon and electron beams dose mapping by means of conventional silver halide films is also common in ion beams. These films are used only for lateral dose distribution measurements being tightly pressed between slabs of plastic, in order to eliminate dose distorting air pockets near the film surface. Due to LET dependence the response of silver halide films exposed at different depths should be converted to dose using film calibrations performed at the correspondent depths employing the same beam type and residual range. Profiles, depth dose and absorbed dose values for small radiosurgery proton beams may be obtained from a single exposure of several radiochromic films placed in the phantom at different depths along beam axis. Radiochromic films are calibrated using a well-characterized uniform radiation field and the dose response curve is obtained in the dose range and conditions of interest. For depth dose measurements films may be also positioned along the direction of the beam but tilted away from beam axis to a small angle to avoid air gap artifacts.

Because of its high quality the alanine dosimetry system can be utilized both for reference and routine proton dosimetry. Measurements of irradiated alanine pellets during one year storage at room conditions did not give any evidence of a long-term alanine fading [00Sha]. This, in addition to the mailing suitability and ruggedness of the dosimeters, may support the possible use of alanine for a transfer dosimetry.

Much research effort has gone into the development of gel dosimetry but this class of 3-D dosimeters has not been yet accepted in clinical practice. One of the reasons is that gel techniques are still very cumbersome. Another important issue is related to the accuracy and precision of the method. Despite the amount of published data, sparse research has been undertaken which provides clear evidence of the accuracy and precision for polymer and Fricke gels at clinically relevant dose levels. In general, gel dosimetry can become a valuable tool in ion therapy as it allows imaging contours of target volumes and dose gradients with high spatial resolution. On the other hand, until the energy- (or LET-) dependent responses of gel dosimeters have been well documented and understood, their acceptance for absorbed dose determination will be limited.

3.4.4.5 Uncertainties

H.-M. KRAMER

According to the Guide to the Expression of Uncertainties in Measurement (GUM) [08ISO] the result of a measurement of a physical quantity should be presented as a combination of the value, its uncertainty and its unit. The requirement applies of course also to dose measurements in teletherapy. Using eq. (3.4.9) as the model equation and applying the steps described in the GUM [08ISO] an uncertainty budget should be established. Examples of uncertainty budgets typical of photon and electron fields are given in DIN 6800 part 2 [08DIN].

3.4.5 Dosimetry for brachytherapy

H.-J. SELBACH

3.4.5.1 Introduction

The term brachytherapy is deduced from the Old Greek word „brachys“, which means short or near. Brachytherapy is a special kind of radiotherapy within the framework of oncology, where in contrast to the teletherapy, the radiation source is placed very close to, in contact of or directly into the tissue, generally the tumour, which is to be treated. Brachytherapy with photon radiation sources has been used successfully in curative treatment of extended macroscopic malignant tumours such as gynaecologic carcinoma, for more than 100 years. The history starts very shortly after the discovery of X-rays by Roentgen in 1895, the discovery of radioactivity by Becquerel in 1896 and of Radium by Marie and Pierre Curie in 1898. Pierre Curie was the first to suggest, in 1901, to use radium in a small tube to be inserted into the tumour, which can be regarded as the start of brachytherapy. Two years later Alexander Graham Bell came up independently with a similar proposal, and in the same year the first successful treatment of cancer with a radium brachytherapy source was reported [07Bal]. Then, several different radioisotopes (e.g. ^{198}Au , ^{60}Co , ^{32}P , ^{125}I) were introduced as brachytherapy sources in the early part of the 20th century. Different applicator systems like the Manchester or Quimby system have been developed in the 1920s and 1930s to achieve an acceptably uniform radiation field within the treatment volume. A more detailed overview over these early developments can be found in [07Bal].

With increasing and extended use of high energy X-rays from linear accelerators in the middle of the 20th century, brachytherapy got into a period of stagnancy or even fell into oblivion. This changed again from the 1980s onward, when new afterloading techniques with high activity sources (^{192}Ir , ^{60}Co) came on the market and radioactive ^{125}I and ^{103}Pd seed implantation became an increasingly popular treatment for localized prostate cancer. In addition beta radiation sources as $^{90}\text{Sr}/^{90}\text{Y}$ or ^{32}P were used for treatments of non-malignant diseases as cardiovascular restenosis after precedent angioplasty. Although this treatment has widely been replaced by the implantation of drug eluting stents, vascular brachytherapy is still used for peripheral vessels.

All in all, modern brachytherapy techniques are playing an important role in the therapy of cancer or non-malignant diseases as a stand-alone radiation therapy or in combination with high-voltage teletherapy.

3.4.5.2 Brachytherapy sources

Table 3.4.4. Properties of radionuclides in currently used clinical photon sources for brachytherapy [00Sch, 11CEA].

Source	Decay	Radiation	$T_{1/2}$ [d]	E_{\max} [keV]	E_{avg} [keV]	Dose-rate
^{60}Co	β^-	γ	1925.23 ± 0.29	1333	1253	LDR, HDR, PDR
^{137}Cs	β^-	γ	10976 ± 29	661	614	LDR
^{192}Ir	β^-, ϵ	$\gamma, (\text{X})$	73.827 ± 0.013	612	371	LDR, HDR, PDR
^{169}Yb	β^-	γ	32.018 ± 0.005	336	93	HDR
^{131}Cs	ϵ	X	9.689 ± 0.016	34.4	32	LDR
^{125}I	β^-, ϵ	$(\gamma), \text{X}$	59.388 ± 0.028	35.5	28	LDR
^{103}Rh	ϵ	X	16.991 ± 0.019	23.2	21	LDR
$^{106}\text{Ru}/^{106}\text{Rh}$	β^-	β	373.6 ± 0.15	3540		

Since the very first treatments of cancer with radium sources, a variety of radioactive sources has been developed and found its clinical application. The different source types can be classified according to the kind of their radiation (β -, γ -radiation), their source strength (high, medium and low dose rate), their energy or application. Furthermore the half-life of the radioactive nuclide is of special interest depending on the application. The most common nuclides employed for brachytherapy sources are listed in [Table 3.4.4](#).

3.4.5.2.1 HDR and PDR sources

Brachytherapy sources delivering dose rates of 0.2 Gy/min and more in 1 cm distance from the source are called high dose rate (HDR) sources according to the definition in ICRU report No. 38. [[85ICR](#)]. The majority of HDR sources are ^{192}Ir sources with initial activities up to 500 GBq, which provides dose rates of the fresh source of about 9 Gy/min at 1 cm distance. This allows short treatment times; even if the source is used for two half lives. In the last decade ^{60}Co HDR sources have also become of major interest, mainly because of their much longer usability due to the longer half live time of 5.3 years compared to the 73.8 days for ^{192}Ir . Because the dose rate constant of ^{60}Co is about three times higher than that of ^{192}Ir , these sources are produced with an initial activity of around 100 GBq, leading to a dose rate of about 5 Gy/min at 1 cm distance. The most common applications of HDR sources are the treatment of tumours of the vaginal apex, oesophagus, lungs, breasts, and prostate.

Pulsed dose rate (PDR) brachytherapy uses sources which are similar to the HDR ones, however, with activities between 20 GBq to 80 GBq (in the case of iridium sources). The treatment is based on short pulses of radiation, typically one per hour, to simulate the overall dose rate of an LDR treatment (see [Sect. 3.4.5.2.2](#)). PDR brachytherapy is mainly used for the treatment of gynaecological, head and neck cancers.

3.4.5.2.2 LDR sources

Low dose rate (LDR) sources are used for permanent or temporary implantation directly into the tumour. They are mainly applied to malignant diseases of the oropharynx, breast carcinoma, sarcoma and, in recent time, in the case of prostate cancer. With the latter application mainly seeds with the radioactive nuclides ^{125}I and ^{103}Pd are used, while for the others ^{192}Ir -wires are frequently taken.

3.4.5.2.3 Ophthalmic applicators



Fig. 3.4.7 Ophthalmic applicator (Eckert & Ziegler BEBIG GmbH)

For the treatment of eye tumours (retinoblastoma and melanoma) ophthalmic applicators as shown in Fig. 3.4.7 are used, which consist of a spherical silver cap containing $^{106}\text{Ru}/^{106}\text{Rh}$ radioactive material completely sealed by the silver surface. $^{106}\text{Ru}/^{106}\text{Rh}$ is a beta emitter with a radioactive half-life of 374 days and a maximum energy of 3.54 MeV. For larger tumours of the eye pure photon emitters like ^{125}I or mixed applicators (electron and photon emitters) are used to permit an irradiation of the tumour at greater depths of the eye than it would be possible with electrons alone.

3.4.5.2.4 Electronic X-ray brachytherapy sources



Fig. 3.4.8 Axxent[®] HDR-X-ray source (Xoft, a division of iCAD).



Fig. 3.4.9 Intrabeam[®] miniature X-ray source (left) with applicator for intra operative radiation therapy (Carl Zeiss Meditec)

In the last decade new developments in miniaturizing of X-ray tubes offered the possibility of using this technique for brachytherapy. Mainly two types of X-ray brachytherapy sources have been developed. One type comprises a very small X-ray tube of less than 2.5 mm in diameter (see Fig. 3.4.8), which is placed within a plastic catheter into the tumour or tissue to be irradiated. High voltage and cooling liquid are supplied via the plastic catheter. The other type of X-ray brachytherapy source consists of a small electron accelerator with a thin evacuated tube of about 20 cm length, acting as an electron beam line, and a gold target at the tip of the tube forming the electron target. A picture of this system is shown in Fig. 3.4.9. Both type of X-ray sources can be operated with accelerating potentials up to 50 keV. The main advantage of the X-ray brachytherapy sources compared to radioactive ones is the possibility of switching the radiation on not before the source is in its radiation position and off at any time.

3.4.5.3 Interstitial brachytherapy

Interstitial brachytherapy represents the oldest form of brachytherapy and encompasses the implantation of radioactive sources, usually in the form of needles, seeds or wires, placed directly into a tumour. Radium needles were the earliest representatives of these sources, which have later been replaced by ^{192}Ir -wires for temporary implantations and by radioactive seeds mainly containing the radionuclides ^{125}I or

^{103}Pd for permanent implantation. While ^{192}Ir -wires are mainly used for the treatment of head and neck, breast, anal, and gynaecological cancers, ^{125}I or ^{103}Pd seeds are nearly exclusively used for the treatment of prostate cancer. On the left of Fig. 3.4.10 two commercial iridium wires 14 cm and 7 cm in length and 0.3 mm in diameter are shown. The picture on the right shows iodine seeds as one example out of some tens of different types of prostate seeds. (see [09Tay] for more types of sources).

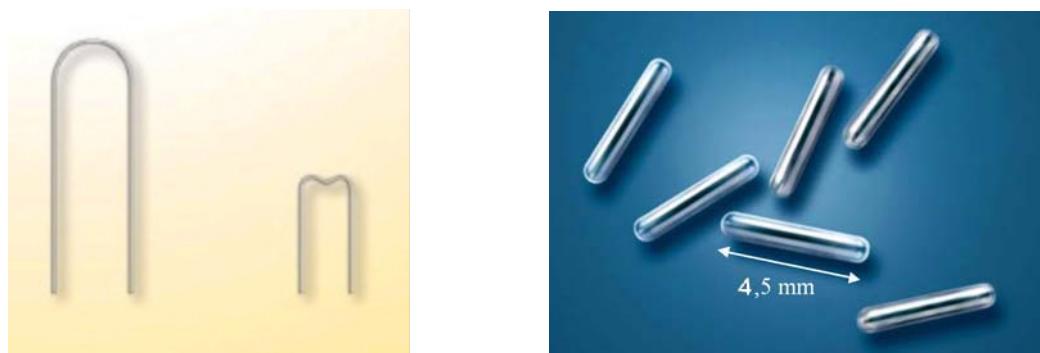


Fig. 3.4.10 Photographies of two types of ^{192}Ir -wires (left) and ^{125}I -seeds (right). (Eckert & Ziegler BEBIG GmbH)

3.4.5.4 Intracavitary brachytherapy

In intracavitary brachytherapy the sources are placed inside a pre-existing body cavity. The most common applications of this method are gynaecological in nature, although it can also be performed on the nasopharynx.

3.4.5.5 Endovascular brachytherapy

In endovascular brachytherapy a catheter is placed inside the vasculature through which sources can be spatially maneuvered. The most common application of this kind is the treatment of coronary in-stent restenosis, although the therapy has also been applied in the treatment of peripheral vascular stenoses and also considered for the treatment of atrial fibrillation. Although a few systems have been used successfully for endovascular brachytherapy, the only device currently available is the Novoste Beta-Cath System from Best Vascular, Inc. which uses beta-emitting sources of $^{90}\text{Sr}/^{90}\text{Y}$.

3.4.5.6 Afterloading technique

Historically, the brachytherapy sources described in the preceding Sections were applied to the patient manually either as temporary or permanent implants. The result of these procedures was that the medical staff performing the treatment was exposed to unacceptable levels of radiation. This problem was solved by the development of remote afterloading systems, which deliver the radioactive materials from a safely contained source via hollow tubes. After the medical staff has placed the catheters and needles according to the pre-calculated treatment plan and after having left the treatment room, the brachytherapy source is remotely driven from the storage container to the pre-defined treatment position. The most frequently used source types for afterloading techniques are ^{192}Ir HDR sources with activities up to 500 GBq producing a rate of absorbed dose to water of about 9 Gy/min at a distance of 1 cm of the source. Because of this initially high dose rate, HDR iridium sources can be used for a time span of up to two half lives (about 150 days, see Table 3.4.4 for exact half live). In recent years ^{60}Co HDR sources of identical dimensions and similar dose rate have been developed, which have the advantage of the longer half live of 5.3 years, so that a source exchange is required only once a decade.

The active material of the majority of HDR afterloading sources has dimensions of 3.5 mm length and 0.6 mm diameter. To be able to irradiate larger target volumes modern afterloading machines are equipped with so called stepping mode and multi-channel features. The stepping mode allows the source to be retracted stepwise in pre-defined steps, thus simulating a line source of arbitrary length, whereas the multi-channel feature offers the possibility of placing (dependent on the type of the machine) 20 needles or even more into the treatment volume and to connect them via catheters with the different channels of the afterloader. The source is then placed automatically in one needle after another, so that greater tumour volumes can be irradiated in a single treatment session.

3.4.5.7 Calibration and traceability

The comparison of the indicated value of a measurement with the value obtained by a standard measuring device is called calibration. Traceability means that there exists an unbroken chain of calibrations from a standard of highest level of accuracy, a primary standard, to the user's measurement device. It should be noted that each additional step in the calibration chain increases the uncertainty of the end-user's measurement. To fulfil the requirement in report 24 of the ICRU [76ICR] that the uncertainty of the dose applied to the treatment volume should not exceed 2.5%, the calibration chain should be kept as short as possible.

A special aspect in the calibration of brachytherapy sources and measurement devices needs to be considered. In radiation therapy the measurement quantity absorbed dose rate to water is worldwide accepted as the relevant quantity on which the treatment of cancer is based. The characterization of the brachytherapy sources, however, are up to now related to air kerma measurements and are traceable to air kerma standards. The quantity in terms of which a brachytherapy source is calibrated is either the reference air kerma rate [04ICR] or the air kerma strength [04AAP], as described in Sect. 3.4.2.4. The quantity absorbed dose rate to water in water at a distance of 1 cm from the centre of the source is obtained by multiplying the reference air kerma rate (or air kerma strength) with the source type specific dose rate constant A . Up to now, mainly Monte Carlo calculations and relative measurement systems are used to determine the conversion of the reference air kerma rate to absorbed dose rate in water at 1 cm from the source. Direct calibrations of sources in terms of absorbed dose rate in water are planned in the near future, but have been only performed twice for the time being [09Bam, 10Sar]. For each specific brachytherapy source type the dose rate constant A has to be evaluated (mainly by Monte Carlo calculations) and published and can be found in special data bases [09Tay, 10EST]

3.4.5.8 Primary standards

3.4.5.8.1 Standards for reference air kerma rate

High Dose Rate sources

With the exception of ^{60}Co HDR-sources, which may be calibrated free in air with graphite ionization chambers evaluated according to the Bragg-Gray principle, common primary standards for the realization of the unit of the air kerma rate can hardly be used for HDR brachytherapy sources for some reasons. High dose rate sources are predominantly based on the radioactive material ^{192}Ir , which spectral distribution covers the photon energy range from about 10 keV to 0.9 MeV with a mean energy of about 370 keV. The broad spectrum of photon energies complicates or even prohibits the usage both established methods for the realization of the unit of the air kerma rate. Neither the principle of secondary electron equilibrium (SEE) (see Sect. 3.2.1.5), which is the prerequisite for measurements with free air ionization chambers up to photon energies of 300 keV, can be used, nor the Bragg-Gray principle (see Sect. 3.2.2.3), on which measurements with small graphite walled cavity ionization chambers are based, could be used in the past without unacceptable uncertainties of the necessary correction factors (mainly the wall correction). Only when Büermann et al. [03Büe] showed the applicability of Monte Carlo calculated correction factors for cavity ionization chambers, it was possible to use a special graphite ionization

chamber, designed as primary standard for ^{60}Co radiation, for $^{192}\text{Ir}-\gamma$ radiation as well [04Sel]. This special parallel-plate ionization chamber with dimensions of 4 cm in diameter and 1 cm thickness and an ionization volume of 6 cm^3 serves at the German National Institute (PTB) as primary standard for ^{60}Co and ^{192}Ir HDR sources with an activity of at least 450 GBq. For sources with lower activities the signal to noise (leakage current) ratio becomes unacceptably large. In these cases the interpolation method described below is used instead.

Due to the difficulties with common air kerma primary standards a transfer standard in combination with an interpolation procedure is widely used. The method was first proposed by Goetsch et al. [91Goe] as a linear interpolation between the calibration factors for an ionization chamber for ^{137}Cs γ -radiation and for a 250 kV X-radiation. At PTB, this method has been improved by measuring the response of a transfer standard, normally an ionization chamber with suitable volume (up to 1000 cm^3), for the X-ray tube voltage range from 10 kV to 300 kV and for ^{137}Cs and ^{60}Co γ -radiation. With the knowledge of the X-ray spectra used the measured response curve is converted to a response curve with respect to monoenergetic radiation, which then can be interpolated at the energy points of the different emission lines of the ^{192}Ir spectrum leading in the end to a calibration factor for a ^{192}Ir radiation field [04Sel, 06Sel].

Low Dose Rate sources

For low dose rate sources like ^{125}I and ^{103}Pd seeds it has also to be realized, that common primary standards for air kerma rate cannot be used, because of the very low reference air kerma rate of some $\mu\text{Gy}/\text{h}$, which does not produce a sufficient ionization current in the limited volumes of the free air chambers used normally for X-radiation. The National Institute of Standards and Technology (NIST) was the first Primary Standard Dosimetry Laboratory (PSDL) in the world to develop a special free air ionization chamber with a sufficiently large volume, the Wide Angle Free Air Chamber (WAFAC) [93Loe, 03Sel], see Fig. 3.4.11. The following description of the WAFAC is taken from the NIST web page (<http://physics.nist.gov/Divisions/Div846/Gp2/wafac.html>).



Fig. 3.4.11 Wide Angle Free Air Chamber.

The Wide-Angle Free Air Chamber (WAFAC) serves as the NIST primary standard for low-energy photon-emitting brachytherapy sources such as ^{125}I . The WAFAC is a variable volume, circular free-air chamber, symmetrical about the beam axis, with lines of force parallel to the beam axis. There is an 80 mm diameter tungsten aperture (on the right of the picture, not visible) at the front of the chamber used to define the beam.

The chamber has three electrodes: the front one (closest to the source) is the polarizing electrode, a middle electrode, and a collecting electrode (farthest from the source and left in the picture) about 150 mm in diameter. The chamber volume is varied from approximately 75 cm^3 to 804 cm^3 by changing the length of middle electrode from 11 mm to 152 mm. The polarizing and collecting electrodes are constructed of aluminized Mylar, about 1 mg/cm^2 thick, while the middle electrode is an aluminum tube 250 mm in diameter. The difference in ionization currents measured with the 152 mm and the 11 mm middle electrode, respectively, is used, along with other parameters, to calculate the air kerma strength of the source.

In Europe the first attempt to develop a primary standard for the realization of the unit of the reference air kerma rate of low-energy interstitial brachytherapy sources and for routine calibrations of ^{125}I and ^{103}Pd LDR sources was made by the German National Metrology Institute, the Physikalisch-Technische Bundesanstalt (PTB). The primary standard developed is a large air-filled parallel-plate extrapolation chamber (GROVEX) with thin graphite front and back electrodes [08Sel]. The chamber is suitable for photon energies up to 40 keV. The underlying principle is that the air kerma rate at the point of measurement is proportional to the increment of ionization per increment of chamber volume at chamber depths greater than the range of the most energetic secondary electrons originating in the entrance electrode. A schematic view of the primary standard is shown in Fig. 3.4.12.

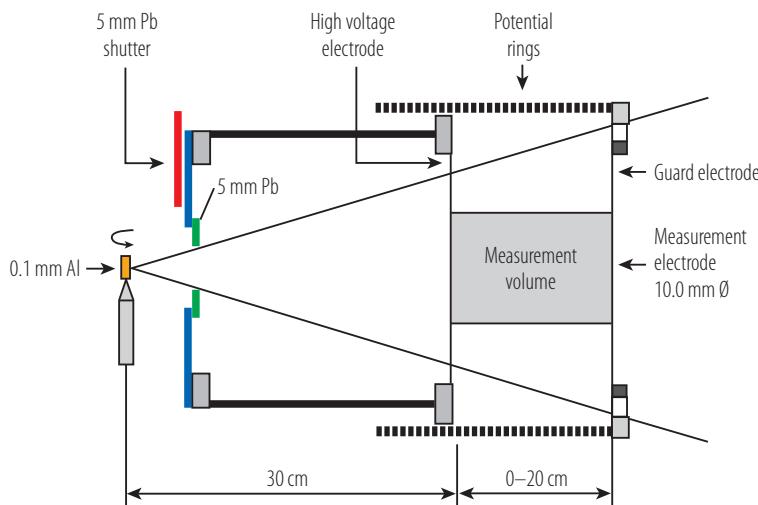


Fig. 3.4.12 Schematic presentation of the large volume extrapolation chamber GROVEX.

The source is mounted within a small aluminium cylinder of 0.1 mm wall thickness, which removes the contaminant titanium K-fluorescence radiation produced in the source encapsulation. A 5 mm thick lead shutter allows background and leakage current measurements with the source in place. A 7 mm thick conical tapered lead aperture located to avoid scattered radiation from the side walls keeps the beam large enough to exceed the diameter of the collecting volume by at least one range of the most energetic electrons produced. A ‘front’ window of graphite diffused onto polyethylene, biased at potential V , acts as the polarizing electrode and serves as the reference plane for the measurement. The ‘back’ window (at ground potential) consists of a centre collecting electrode and an outer guard ring, both consisting of graphite diffused onto polyethylene. The collecting electrode serves as the defining cross-sectional area for the measuring volume and was diffused with a nominal diameter of 100 mm. The reference point for determining \dot{K}_δ is taken at the centre of the reference plane, which is located 30 cm from the source.

The reference air kerma rate, \dot{K}_δ , by means of the extrapolation chamber technique is given by

$$\dot{K}_\delta = \frac{(W/e)_{\text{air}}}{\rho_{\text{air}} A_{\text{eff}} (1 - g_{\text{air}})} \frac{d(kI)}{ds} \prod_i k_i \quad (3.4.35)$$

where $(W/e)_{\text{air}} = 33.97 \text{ eV}$ is the mean energy required to produce an ion pair in air, $\rho_{\text{air}} = 1.2046 \text{ kg/m}^3$ is the density of air at 20°C and 1013.25 hPa. A_{eff} is the effective area of the measuring electrode, $g_{\text{air}} \approx 0$ (for energies less than 100 keV, and see Sect. 3.2.1.5) is the fraction of the initial electron energy lost by bremsstrahlung production in air, $d(kI)/ds$ is the increment of corrected ionization current (background and leakage subtracted, corrected to reference temperature and pressure) per increment of the chamber volume and k_i are corrections to the measured currents. The derivative $d(kI)/ds$ represents the slope of the charge–volume relationship.

3.4.5.8.2 Standards for absorbed dose rate to water

As described in the previous chapters, the quantity realized by the existing primary standards is the reference air kerma (rate). However, the therapy is based on the quantity absorbed dose (rate) to water. Therefore dose calculation protocols and published data for the source used are needed to convert the reference air kerma rate into the absorbed dose rate to water in 1 cm distance from the source. However, most of the calculated and published data for the conversion to absorbed dose rate to water have a quite great uncertainty (5% to 8%). The preferable alternative to the air kerma calibrations is to calibrate the sources directly in the desired quantity, the absorbed dose to water in 1 cm distance from the source. Several National Metrology Institutes are therefore developing primary standards for the realization of the unit of the absorbed dose rate to water in the vicinity of the source.

HDR sources

For high dose rate sources calorimetric measurements are the method of choice, where graphite calorimeters as well as water calorimeters are being employed. For the time being (spring 2011) two new graphite calorimeters are being constructed at the National Physics Laboratory (NPL) in London, UK, and at the Italian National Institute (ENEA) in Rom. Furthermore calibrations of ^{192}Ir HDR sources in terms of absorbed dose to water have already been performed by means of a water calorimeter at two metrology institutes, at the Physikalisch-Technische Bundesanstalt (PTB) in Braunschweig, Germany [09Bam] and at the National Research Council of Canada (NRC) in cooperation with the McGill University, Montréal General Hospital, Montréal respectively [10Sar]. A third water calorimeter for ^{192}Ir HDR sources is just being taken in operation at the Dutch metrology institute, the Van Svinde Laboratorium (VSL) in Delft. The design of the water calorimeters used for the measurements with HDR brachytherapy sources is mainly the same as of the primary standard water calorimeters operated in a ^{60}Co beam. The calorimeters are also operated at a water temperature of 4°C to avoid convection in the calorimeter and similar types of detectors are used. However, some modifications have been made to position the HDR source at a short distance in front of the detector inside the water phantom. Fig. 3.4.13 shows as an example the modified detector unit of the PTB calorimeter. A stainless steel needle is fixed in front of the detector unit, a high purity water filled glass cylinder with two sensors, and connected to a polytetrafluorethen (PTFE) catheter, through which the source is positioned into the tip of the needle in 25 mm distance from the temperature sensors.

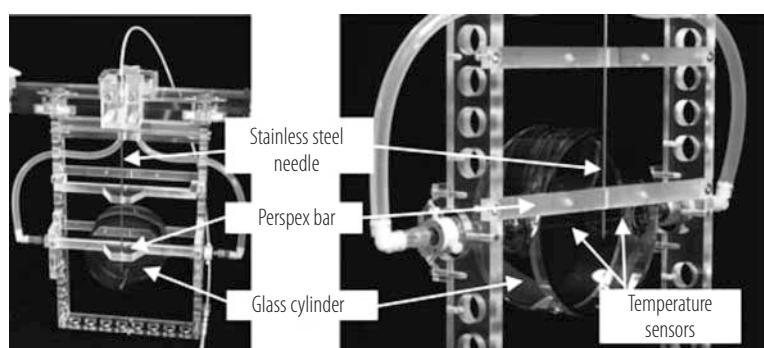


Fig. 3.4.13 Picture of the calorimetric detector within its frame structure, also showing the additional bars to fix the stainless steel tube at both sides of the detector.

The main problem associated with the calibration of HDR sources is the short distance between source and detector, which causes a pronounced divergent radiation field and a steep dose rate gradient at the point of the detector due to the $1/r^2$ law. An additional problem arises from the source itself, because due to the absorption of electrons and photons in the source itself it also acts as a heat source. The heating power for a 500 GBq ^{192}Ir source, for example, amounts to about 30 mW. Dependent on the distance between source and detector and dependent on the irradiation time the self-heating influences significantly the necessary heat conduction corrections. Fig. 3.4.14 compares the results of calorimetric measurements in an external ^{60}Co beam with 90 s irradiation time with measurements of the temperature increase in the vicinity of a ^{192}Ir source for three different irradiation times. The influence of the heat

wave caused by the self-heating power can clearly be seen. This effect is corrected by complex heat transfer calculations on the basis of Finite Element Methods (FEM) [09Bam].

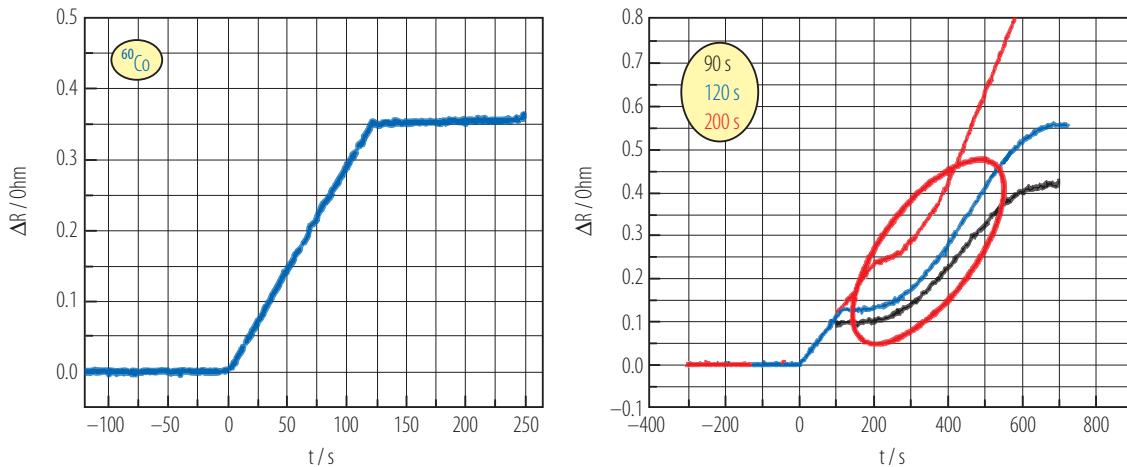


Fig. 3.4.14 Measured temperature increase characterized by resistance change during and after 90 s irradiation for an external ^{60}Co radiation field. (left), and for three different irradiation times and an irradiation field of a ^{192}Ir source in 25 mm distance from the detector (right). A resistance change of 0.1 Ohm represents a temperature increase of about 2.5 mK or an absorbed dose to water of 10.4 Gy

LDR sources

The ^{125}I and ^{103}Pd seeds as used for interstitial brachytherapy usually have dose rates of 10 mGy/h to 50 mGy/h in 1 cm distance from the source. These low dose rates exclude the calorimetric measurement of the absorbed dose to water. A dose rate of 50 mGy/h will cause a temperature rise of only 12 μK after 1 h irradiation time, which is impossible to measure. Therefore, similar large volume extrapolation chambers as those for the realization of the unit of the reference air kerma rate are used. The difference to the air kerma extrapolation chambers is, that these standards are equipped with a 1 cm water-equivalent front electrode and a water-equivalent phantom at the back electrode, allowing in-phantom (in-water) extrapolation measurements. As phantom material water-equivalent plastic is used (PTB) as well as graphite (ENEA). A different approach is done by the French metrology institute (LNHB) where air kerma measurements around a 2 cm in diameter water-equivalent sphere with the source in the centre are performed and the conversion into absorbed dose to water is done by Monte Carlo Calculations. For all the three approaches of the realization of the unit of the absorbed dose rate to water, the measurements procedures are not as direct as the calorimetric measurement performed with dose rate sources, so that extensive MC simulations are necessary in all three cases.

3.4.5.9 Spatial distribution of dose rate

Although both LDR and HDR brachytherapy sources are very small they cannot be considered as point sources with an isotropic distribution of the radiation, which would imply that isodose lines or planes have exact circular or spherical shape respectively. The majority of the sources are of similar geometry, having a length between 3 mm and 5 mm and a diameter around 0.5 mm. In the vicinity of the sources 20 mm and closer this leads to a geometrical correction discussed in more detail in the next Section. Another reason for a non-isotropic dose rate distribution is caused by the manufacturing of the source encapsulation; normally a thin stainless steel or titanium tube with soldered end caps at both ends and in the case of HDR sources attached with one end to a metal guide wire. Inhomogeneous distribution of the activity, differences in the thickness of the tube and in the end caps result in a more or less pronounced anisotropy of the dose rate around the source perpendicular to source axis (radial or azimuthal anisotropy)

as well as in the direction of the source axis (polar anisotropy). In Fig. 3.4.15 examples of the radial (left) and the polar (right) anisotropy of a ^{125}I seed are shown.

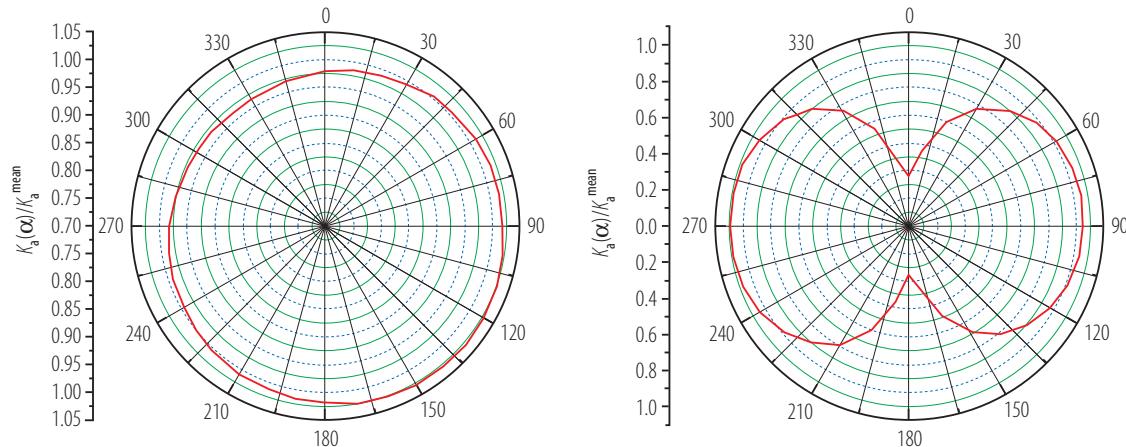


Fig. 3.4.15 Radial distribution of the air kerma rate (left), and polar distribution of the air kerma rate (right), normalized to the mean reference air kerma rate

The diagrams illustrate some of the difficulties associated with the dissemination of the measurement quantity to the end user. The reference air kerma rate as well as the absorbed dose rate to water in 1 cm are defined in a point perpendicular to the source axis. The transfer standard normally used is a well type chamber, which has a nearly isotropic response over the whole solid angle of 4π . The radial anisotropy can be accounted for in the source characterization by rotating the source and thus averaging the air kerma rate over the 360° angle. The polar anisotropy, however, causes different calibration factors for the well type chamber for different polar dose rate distributions. Therefore the concept of transferring the air kerma rate or absorbed dose rate in a point at a certain distance from the source by means of well type chambers requires a special calibration factor for each different type of source (even for the same nuclide) and negligible variation of the anisotropy from one source of a given type to another. Otherwise the uncertainty of the calibration becomes unacceptably large or the whole concept of characterizing the source by its dose rate in a point perpendicular to its axis becomes questionable.

3.4.5.10 Dose calculation and protocols

Several protocols for source handling, calibration and dose calculation have been produced by national and international institutions to provide practical help for the medical physicist in the hospital. Beside the Code of Practice from the IAEA [02IAE], the reports of the American Association of Physicists in Medicine (AAPM) have to be mentioned [04AAP, 78AAP, 97Nat]. Although a quite old German Standard, the DIN 6809-2 exists [93DIN], which recommends the calibration in terms of absorbed dose to water, most of the protocols are based on the calibration of sources in terms of the reference air kerma. A world wide accepted protocol today is the report of Task Group 43 (TG43) of the AAPM [04AAP]. This report explicitly addresses LDR sources, but the dose calculation algorithm is applied to HDR-sources as well.

The basic concept of the dose calculation according to TG43 is briefly described in the following using the geometry shown in Fig. 3.4.16.

The absorbed dose rate to water $\dot{D}(r, \theta)$ at a point $P(r, \theta)$ at a distance r from the source centre at the polar angle θ relative to the source longitudinal axis is given by the formula:

$$\dot{D}(r, \theta) = S_K \Lambda \frac{G_L(r, \theta)}{G_L(r_0, \theta_0)} g_L(r) F(r, \theta). \quad (3.4.36)$$

S_K is the air kerma strength, which is numerically identical to the quantity reference air kerma rate which is used instead in this work (see Sect. 3.4.2.4) and recommended by the ICRU [04ICR]. The conversion factor Λ (called dose rate constant in TG43) converts the reference air kerma rate to the absorbed dose rate to water at the reference point $P(r_0, \theta_0)$ at 1 cm distance from the source centre. $G_L(r, \theta)$ is a geometry function, which accounts for the differences in the dose rate distribution between an ideal point source and a real source of length L . In the case of the point source approximation $G_L(r, \theta)$ becomes $1/r^2$ independent of the angle θ . For a source length $L \neq 0$ the geometry function is given as:

$$G_L(r, \theta) = \frac{\beta}{Lr \sin \theta}, \quad \text{if } \theta \neq 0^\circ, \quad (3.4.37)$$

and

$$G_L(r, \theta) = (r^2 - L^2/4)^{-1}, \quad \text{if } \theta = 0^\circ, \quad (3.4.38)$$

where β can be taken from the coordinate system shown in Fig. 3.4.16. For commonly used brachytherapy sources with a length of the radioactive core of 4 mm or less the geometry function differs significantly from the value 1.0 only for distances below 2 cm.

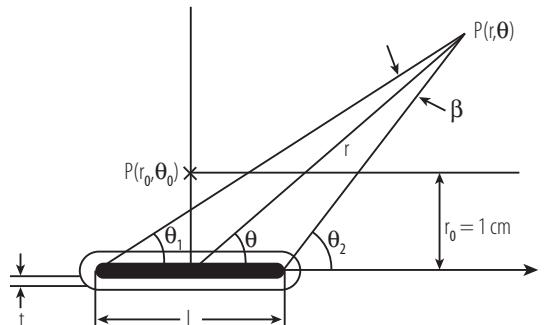


Fig. 3.4.16 Coordinate system used for brachytherapy dosimetry calculations (taken from the AAPM report TG43).

The radial dose function, $g_L(r)$, represents the relative depth dose curve perpendicular to the source axis multiplied by the geometry function $G(r, \theta = 90^\circ)$. This function accounts for photon scattering and attenuation. $F(r, \theta)$ is the two-dimensional anisotropy function which describes the variation in dose as a function of polar angle relative to the transverse plane, as described in Sect. 3.4.5.9. All three functions $g_L(r)$, $G_L(r, \theta)$ and $F(r, \theta)$ are normalized to unity at the reference point $P(r_0, \theta_0)$, which means, that at this point the absorbed dose rate to water is given by the product of reference air kerma rate (or air kerma strength) and the dose rate conversion factor Λ . This concept of dose calculation presumes that the factor Λ and the functions $g_L(r)$, $G_L(r, \theta)$ and $F(r, \theta)$ are source type specific and are - within acceptable uncertainties - identical for each source of a given type. For all source types used for brachytherapy consensus data for Λ , $g_L(r)$, $G_L(r, \theta)$ and $F(r, \theta)$ are published and available on the world wide web as for example at the ESTRO database [10EST] or from the Carleton University, Canada, [09Tay] and for some source types reference data are given in TG43 [04AAP].

3.4.5.11 Uncertainties

The uncertainty of the dose delivered to the patient is comprised of several components, starting from the realization of the unit of the measurement quantity, which, for the time being, is the reference air kerma rate, the calibration of a transfer standard, the determination of the reference air kerma rate for a specific source at the hospital, the dose conversion factor Λ , the dose planning system or algorithm, and last but not least the correct positioning of the source in the treatment volume during the medical application. For the two last contributions one has to rely on rather crude estimates as confirmed investigations on the accuracy of dose planning systems in brachytherapy as well as data on geometrical uncertainties in source

positioning are rare or missing altogether. Some estimates on uncertainties of dose calculations can be found in the report TG43 [04AAP].

The relative standard uncertainty ($k = 1$) for the realization of the unit of the reference air kerma rate for HDR ^{192}Ir sources is between 1% and 1,5% dependent on the metrology institute. Currently only the National Institute for Standardization and Technology NIST and the Physikalisch-Technische Bundesanstalt PTB operate a primary standard for LDR sources with a standard uncertainty of about 1% ($k = 1$). Uncertainties of the realization of the unit of the reference air kerma rate are to be found in the Calibration and Measurement Capability (CMC) list of the Bureau International des Poids et Mesures (BIPM). [09BIP]. The uncertainties given above and listed in the CMC list are the lowest uncertainties possible. Due to source dependent parameters as insufficient dose rate, homogeneity of the activity distribution, anisotropy etc, these uncertainties may be significantly larger. In addition the conversion factor A is associated with a non negligible uncertainty dependent on the method of evaluation of this factor for a specific source type. The uncertainty of A can be in the order of 5% ($k = 1$). The efforts of the metrology institutes in developing primary standards for the quantity absorbed dose to water are aimed at a significant reduction of this uncertainty by direct calibration of transfer standards in terms of absorbed dose to water.

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4.1 Introduction

Dosimetry in Nuclear Medicine Diagnosis and Therapy

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The use of small amounts of radioactive materials (or tracers) to help diagnose a variety of diseases is called diagnostic nuclear medicine or more recently functional and molecular imaging. The radionuclides used are either in ion form or more often attached to a ligand with specific affinity for a physiological process, a receptor, an antigen or a protein. The penetrating gamma radiation (sometimes characteristic X-rays and even bremsstrahlung) is registered in an external detector most often in the form of a gamma camera or positron emission tomography (PET) camera. Using higher activity and radionuclides with suitable characteristics (such as β -, Auger electron or α -emitters; see Sect. 2.2.1), the technique is used for radionuclide therapy. An element common to all nuclear medicine is the need for radiation dosimetry. It is fundamental to the safety of the patients undergoing diagnostic nuclear medicine. Still higher accuracy in the absorbed dose estimates is needed for patients undergoing radionuclide therapy in order to guarantee the therapeutic outcome and to minimize adverse effects in normal organs and tissues.

Advances in our understanding of the molecular biology of cancer and other diseases have identified new molecules and signaling pathways that we can now visualize, *in vivo*, for diagnosis, staging, to identify optimal therapy, to carry out the therapy and to monitor patient response to therapy.

4.1.1 Why dosimetry in nuclear medicine?

As the initial step in the interaction of ionizing radiation with matter is energy transfer, which leads to ionizations, it might be reasonable to use – at least as a first approximation – the amount of absorbed energy per unit of mass (the absorbed dose) to quantify the radiation exposure in order to estimate the effect caused by a given exposure. For more information about possible biological effects, the reader is referred to Chapter 2 “Biological Effects of Ionising Radiations” in Landolt-Börnstein Vol. VIII/4 “Radiological Protection” [05Kau], and to Sect. 2.3 “Biological Effects” in the present Volume.

Internal dosimetry deals with the determination of the absorbed dose and its spatial and temporal distribution from the decay of radionuclides in organs and tissues in the body. For more detailed information the reader is recommended to read Chapter 7 “Internal Dosimetry of Radionuclides” and Chapter 4 “Radiological Quantities and Units” in LB Vol. VIII/4 [05Kau], and Chapter 3 “Dosimetry in Diagnostic Radiology and Radiotherapy” in the present Volume.

In diagnostics, sufficiently accurate absorbed dose estimates for an average patient for each specific investigation is needed in order to optimise the use of the various alternative radiodiagnostic techniques and to be able to estimate the collective radiation exposure and risk from nuclear medicine of developing radiation-induced cancer later in life and also to make possible population dose estimates and judge if the benefit of an examination justifies its risk. Together with judgement of the diagnostic value/image quality, a dose estimate is an essential parameter in the optimisation of diagnostic procedures for various groups of patients and investigations.

The use of radiopharmaceuticals for therapy requires even more detailed and also patient- and tissue-specific dosimetry and dose planning. Radiotherapy with radiopharmaceuticals requires individual dose estimates for the tumour as well as for normal tissues (bone marrow, liver, kidneys, etc.) to predict tumour effect and normal tissue acute and late reactions. In many situations the tolerance dose of normal tissues limits the dose that can be given to the tumour. Even if the therapeutic use of radiopharmaceuticals is small today compared to the diagnostic use, developments of methods for internal dosimetry are being pushed forward by an interest in improving the dosimetry in connection with therapy. The result of this development is also of interest for the dosimetry in diagnostic nuclear medicine.

Reliable dosimetry is also needed to check that the exposure of radiation workers does not exceed the dose limits and that their exposure is kept as low as possible. This is dealt with in more detail in LB Vol. VIII/4 [05Kau], Sections 10.3.2 and 10.3.3, and in the present Volume in Sects. 5.3 and 5.4. The occupational exposure is normally dominated by external radiation, which is supervised using personal dosemeters. The internal exposure has to be checked by measurements of the body burden of gamma-emitters and through analysis of excreta for analysis of beta- and alpha-emitters.

4.1.2 Relation between absorbed dose and biological effect

All work on radiation dosimetry is based on the assumption that the absorbed dose in a tissue or organ is a good predictor of effects at least at certain dose levels and dose rates. In external radiation therapy there are many convincing examples of very clear correlations between absorbed dose and effects. The same can be assumed for radionuclides if they are uniformly distributed in an organ or tissue. However, it may not necessarily be a good predictor of response when the activity is not uniformly distributed and when the energy is deposited by short-range particles like Auger-electrons. For α -particles, which have a much longer range than Auger electrons, the effect per unit absorbed dose is higher than for electrons (photons) due to the higher LET. Further work will show if the implementation of radiobiological models and calculation of a biologically effective dose (BED) will improve the prediction possibilities (see Sect. 4.10.2).

4.1.3 Absorbed doses cannot be measured *in-vivo*

As the absorbed dose to various organs and tissues in the body cannot be measured, biokinetic and dosimetric models (see Sect. 4.2.4) are needed to calculate conversion coefficients, which link measurable quantities like administered activity and kerma rate in air to absorbed dose in tissues and organs.

4.1.4 Historical background

During the development of diagnostic and therapeutic nuclear medicine there have been increasing efforts to quantify the exposure of various organs and tissues of the investigated or treated individual. The development and the current status of medical internal dosimetry has been described in a number of review papers, books and book chapters [70Qui, 95Ste, 99Mat, 00Zan, 04Sgo, 05Hub, 07Sta, 08Sta1, 08Sta2].

Radiation dosimetry for patients began to develop gradually as artificial radionuclides became more easily available in the late 1930's. ^{32}P was used as a substitute for whole-body X-ray treatments at various blood disorders like leukaemia and *polycytemia vera*. About the same time the value of ^{131}I -iodide in the study of thyroid function was demonstrated and within some years ^{131}I -iodide was also used in the treatment of thyrotoxicosis and thyroid cancer. ^{59}Fe was simultaneously used in various studies relating to red blood cells and iron reserves.

At that time, techniques for internal dose calculations existed for therapeutic implants of radium (^{226}Ra) and radon (^{222}Rn) in capsules or needles. Different techniques, source arrangements and dosimetric systems were developed. Some of the first attempts to develop a system for internal dosimetry for radiopharmaceuticals were made by Marinelli [42Mar]. At that time only the internal absorbed dose delivered by beta particles was estimated. This was considered to be acceptable as it was several times higher than the absorbed dose contribution due to photons from the radionuclides in use at that time. Marinelli also noted that the main difficulty in internal dosimetry is the lack of data related to the metabolic and biological elimination of the substances. After the Second World War, Marinelli, Quimby

and Hine [48Mar, 51Qui1, 51Qui2, 58Qui] further developed the pre-war dosimetry and also addressed the problem of calculating the radiation dose from gamma emitters. For the gamma-rays, the basic information became the specific gamma (ray) constant (Γ). This was defined as the exposure in röntgen (R) at 1 cm distance from a naked point source of 1 mCi in air during 1 hour. Quimby noted that many organs could be approximated by spheres, a concept later used by ICRP in its early report on internal dosimetry [59ICR, 71ICR]. Others put cylinders together to simulate a human body. If a gamma emitter is distributed uniformly in an organ (or phantom), the absorbed dose will not necessarily be uniform, and it depends on the position of the point of interest. At any point it will be the summation of dose contributions from all volume elements and thus it will depend on the geometry and of the absorption of the radiation. Starting with Γ , the absorbed dose at a point from an infinitesimal volume element is integrated over the entire source volume to get the absorbed dose in the point. This integral is called the geometric factor, g , for the point.

Also Quimby pointed out that human data on the distribution of radionuclides were lacking and that data on long-lived radionuclides were especially scanty. Marinelli's concept of treating low-energy gamma and x rays as if they were beta radiation [48Mar] was extended by Loevinger et al. [56Loe] to include other forms of non-penetrating radiations such as internal conversion electrons, Auger electrons, and low-energy x rays.

Based on this early internal dosimetry and available biokinetic information dose data for "typical" patients were calculated and summarised in practically useful tables [58Vea, 59Gar, 69Gar, 73Kau].

A great step forward was taken in the 1970s with the establishment of the Medical Internal Radiation Dose (MIRD) Committee of the Society of Nuclear Medicine in USA and the development of the "MIRD phantoms" and calculation techniques [69Sny, 75Sny, 78Sny, 91Loe]. Much of the internal dose calculation have then been based on work done by the dosimetry research group at Oak Ridge National Laboratory (ORNL) in USA (Snyder, Ford, Watson, Schlafke-Stelson, Cloutier, Coffey and others). Based on ICRP Publication 23, a massive collection of anatomical and physiological data that defined ICRP's "Reference Man", Snyder and co-workers formulated a 3D model of the body, incorporating Monte Carlo methods for modelling radiation transport, to calculate absorbed dose to various organs and tissues. The work done at Oak Ridge and other places in the world was continuously followed up at the very popular and informative Oak Ridge International Symposia on Radiopharmaceutical Dosimetry (1970–2002) which are now being continued as special sessions at the annual congress of the European Association for Nuclear Medicine or of the Society of Nuclear Medicine, USA. The new dosimetric methods and new biokinetic data were used by Kaul, Roedler and Henrichs [79Kau] in Germany, Veall and Smith in England and Nosselin, Johansson and Mattsson [78Joh] in Sweden to produce the ICRP Publication 53 on "Radiation dose to patients from radiopharmaceuticals" [87ICR]. ICRP has since then produced a number of addenda, e.g. [98ICR] and [08ICR], to this publication.

The work at Oak Ridge was later focused on the development of age- and gender-specific dose estimates. This work was mainly motivated by the interest for dose estimated for occupationally exposed persons and for members of the public, but has also been of much value for the medical dosimetry. As a first step, Cristy and Eckerman extended the radiation transport models and mathematical phantoms from a reference adult male to adult females and children [87Cri].

In diagnostic nuclear medicine, internal absorbed dose calculations are normally carried out using the MIRD scheme (see Sect. 4.2.2.1) which is being replaced by the joint MIRD/ICRP scheme described in Sect. 4.2.2.3. The calculations require residence times for radiopharmaceuticals and S -values for organs of interest. Residence times are obtained using the clinical imaging equipment. S -values are built into the practically very useful calculational tools MIRDOS, which contains S -values for over 200 radionuclides [96Sta] and OLINDA¹/EXM, which contains S -values for a large number of radionuclides [05Sta]. Both programs are based on the MIRD stylised mathematically describable phantoms (adult male and female, child and pregnant woman).

¹ OLINDA – Organ Level INternal Dose Assessment

4.1.5 References for 4.1

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4.2 Methods for Internal Dosimetry

Dosimetry in Nuclear Medicine Diagnosis and Therapy

D. NOBKE, S. MATTSSON, L. JOHANSSON

4.2.1 General assumptions and definitions. Quantities and units

The method of dose calculation described here is basically the same as in [Section 4.7](#) of Landolt-Börnstein Vol. VIII/4 [[05Kau](#)] but another terminology is used here. The reason is that the terminology of [[05Kau](#)] is that used by ICRP [[77ICR](#)] for internal doses of both workers and members of the public, see [Sect. 4.2.2.2](#), while ICRP in its Publications 53, 80, and 106 [[87ICR](#), [98ICR](#), [08ICR](#)] used the MIRD terminology [[68Loe](#), [91Loe](#)] for the calculation of internal doses to patients from radiopharmaceuticals, see [Sect. 4.2.2.1](#). To avoid confusion MIRD and ICRP determined a unified internal dosimetry methodology [[09Bol](#)], see [Sect. 4.2.2.3](#), which will be used by MIRD as well as in forthcoming ICRP Publications and which is used here.

4.2.1.1 Organ doses

The initial step in the interaction of ionizing radiation with matter is energy transfer which leads to ionizations. It might be reasonable to use the amount of absorbed energy per unit of mass to quantify the radiation exposure in order to estimate the effect caused by a given exposure.

For assessing doses from radiation exposures special dosimetric quantities have been developed by the International Commission on Radiation Units and Measurements (ICRU) and the International Commission on Radiological Protection (ICRP). The absorbed dose is the basic physical dose quantity. It is used for all types of ionizing radiation and any irradiation geometry (for further information see LB Vol. VIII/4 [[05Kau](#)], Chapter 4, and Chapter 3 in the present Volume).

The absorbed dose, D , is defined as the quotient of $d\bar{\epsilon}$, by dm , where $d\bar{\epsilon}$ is the mean energy imparted to matter of mass dm by ionizing radiation, that is

$$D = \frac{d\bar{\epsilon}}{dm} \quad (4.2.1)$$

The SI unit is J kg^{-1} and its special name is gray (Gy).

In radiation protection in general, absorbed dose averaged over the target region r_T , either a target organ (as liver or kidneys, for example) or a target tissue (for example muscle), $D(r_T)$, is considered. In some cases averaged doses for radiosensitive regions of an organ or tissue instead of doses for the whole organ or tissue are considered. Examples for this are the walls of the human respiratory tract [[91ICR](#)] and the human alimentary tract [[06ICR](#)], the skeleton (consideration of a 10 μm thin layer at the surface of the bone [[77ICR](#)]), and the skin (a thin layer in a depth of 70 μm from the skin surface [[77ICR](#)]).

Some types of radiation are biologically more effective than others. This is reflected by a radiation weighting factor, w_R , and by the introduction of the term equivalent dose. The equivalent dose to a target organ or target region r_T , $H(r_T)$ is defined by

$$H(r_T) = \sum_R w_R \cdot D_R(r_T) \quad (4.2.2)$$

where $D_R(r_T)$ is the mean dose to the target organ or tissue r_T due to radiation type R . Because the radiation weighting factor w_R is dimensionless the SI unit of the equivalent is also J kg^{-1} , but its special name is sievert (Sv).

Due to the latest ICRP recommendations [[07ICR](#)] the radiation weighting factor is 1 for photons, electrons, positrons, and β particles, and 20 for α particles. Some radionuclides used in nuclear medicine

(e.g. ^{99m}Tc , ^{123}I , ^{125}I , and ^{201}Tl) emit Auger electrons. The biological effects of these low-energy electrons may be higher than that of electrons and β particles when the radionuclide is incorporated into the DNA of the cell nucleus. ICRP acknowledges that for Auger electron emitters bound to DNA a larger radiation weighting factor (of about 20) may be appropriate but it does not recommend a specific value. The American Association of Physicists in Medicine (AAPM) report no. 49 [94H1] recognized that the RBE caused by Auger electrons emitted by DNA-incorporated radionuclides is similar to that seen for α particles and, thus, has recommended a radiation weighting factor of 20 for Auger electrons for stochastic effects and of 10 for deterministic effects. Furthermore, the AAPM has recognised that the equivalent dose for Auger electron emitters is dependent on that subcellular distribution.

In internal dosimetry, the committed equivalent dose, the equivalent dose delivered over a defined period, T_D , is considered:

$$H(r_T, T_D) = \int_0^{T_D} \dot{H}(r_T, t) dt \quad (4.2.3)$$

where $\dot{H}(r_T, t)$ is the equivalent dose rate in the target organ or tissue r_T at time t . In general T_D is taken to be 50 years for adults and the time until the age of 70 years for children or adolescents. Because of the rather short half-lives of the radionuclides used in nuclear medicine T_D is taken to be infinite in those applications.

4.2.1.2 Effective dose

The effective dose was introduced in ICRP Publication 26 [77ICR] as a measure for the risk to get a radiation-induced probabilistic detriment like cancer and hereditary effects in an occupationally exposed population. It is a weighted mean value of the tissue doses with tissue weighting factors w_T . The tissue weighting factors of ICRP Publication 26 have been revised in ICRP Publication 60 [91ICR] and ICRP Publication 103 [07ICR].

In the methodology of ICRP Publications 26 [77ICR] and 60 [91ICR] the effective dose E is

$$E = \sum_T w_T \cdot H(r_T) \quad (4.2.4)$$

The tissue weighting factors of ICRP Publication 60 are listed in column 2 of [Table 4.2.1](#). The remainder dose is the dose averaged over 12 listed organs and tissues: Adrenals, brain, extrathoracic airways, small intestine, kidneys, muscle, pancreas, spleen, thymus and uterus. If, however, the highest tissue dose is the dose to one of the 12 remainder organs and tissues listed above then the splitting rule applies: half of the tissue weighting factor for the remainder (i.e. 0.025) is assigned to that highest tissue equivalent dose and the other half is assigned to the other remainder tissues.

Table 4.2.1. Tissue weighting factors w_T of ICRP Publication 60 [91ICR] and ICRP Publication 103 [07ICR].

Organ	ICRP Publication 60	ICRP Publication 103
Gonads	0.20	0.08
Red bone marrow	0.12	0.12
Colon	0.12	0.12
Lung	0.12	0.12
Stomach	0.12	0.12
Urinary bladder	0.05	0.04
Breast	0.05	0.12

Organ	ICRP Publication 60	ICRP Publication 103
Liver	0.05	0.04
Oesophagus	0.05	0.04
Thyroid	0.05	0.04
Skin	0.01	0.01
Bone surface	0.01	0.01
Brain	-	0.01
Salivary glands	-	0.01
Remainder	0.05	0.12

ICRP Publication 103 has changed this concept a little bit. To reflect that the concept of effective dose has to be applied to large groups of population of all ages and both gender rather than to individuals the sex-averaged organ equivalent doses for the reference male and reference female are used for its calculation:

$$E = \sum_T w_T \cdot \frac{(H(r_T)^{\text{Male}} + H(r_T)^{\text{Female}})}{2} \quad (4.2.5)$$

In this new formalism the remainder dose is the arithmetic mean of the dose to 13 organs and tissues: Adrenals, extrathoracic airways, gallbladder, heart, kidneys, lymphatic nodes, skeletal muscle, oral mucosa, pancreas, prostate (males), small intestine, spleen, thymus, and uterus/cervix (females). There is no splitting rule any more, and therefore the first time in its history the effective dose is (formally) additive.

Following the ICRP Publication 103, the effective dose must be calculated using the ICRP reference voxel models (for adults published in ICRP Publication 110 [09ICR]) and (for internal emitters) the latest ICRP biokinetic models adequate for the situation and substance considered. In nuclear medicine it may be used for comparing doses from different diagnostic procedures and for comparing the use of similar technologies and procedures in different hospitals and countries as well as the use of different technologies for the same medical examination. However, it must not be used for planning the individual exposure of patients and individual risk-benefit assessments; for this the equivalent dose or the absorbed dose to irradiated tissues is the relevant quantity to be used.

4.2.2 Dosimetry systems

Absorbed dose estimates from radiopharmaceuticals require knowledge about *biokinetic data* (e.g. cumulated activities in organs and tissues) as well as *physical data* (e.g. decay properties of the radionuclide, emitted radiation and source-target geometry).

The history of the development of dosimetry systems starting with sphere geometries [48Mar] has been outlined in Sect. 4.1.4. Here we start to describe the first more advanced dosimetry system, the MIRD system, see Sect. 4.2.2.1, which has been developed for dose calculations in nuclear medicine. This system is similar to the system used by ICRP for internal dose assessment for workers and members of the public see Sect. 4.2.2.2. At the end a synthesis of both systems will be shown, see Sect. 4.2.2.3, which will be used by MIRD and ICRP in the future.

Traditionally, the mean absorbed dose to an organ or tissue (even to the whole body) – the so-called macroscopic dose is considered. To better describe the biological effects and risks, it might be an advantage to also carry out small-scale dosimetry (for different tissues in an organ, as for example the latest approaches for skeleton dosimetry, Sect. 4.2.3.2) and even dosimetry on the cellular and sub-cellular level (microdosimetry).

4.2.2.1 MIRD system

In 1976, the Medical Internal Radiation Dose (MIRD) Committee published a schema for the calculation of doses from incorporated radionuclides [76Loe] based on an older method [68Loe].

In this schema the absorbed dose to a target region r_k is calculated by

$$\bar{D}(r_k) = \sum_h \tilde{A}_h \cdot S(r_k \leftarrow r_h) \quad (4.2.6)$$

where \tilde{A}_h is the cumulated activity in the source region r_h , [Bq·s], and $S(r_k \leftarrow r_h)$ is the mean absorbed dose in target region r_k per unit cumulated activity in source region r_h , [Gy·(Bq·s) $^{-1}$].

The cumulated activity is calculated by

$$\tilde{A}_h = \int_0^{\infty} A_h(t) dt \quad (4.2.7)$$

where $A_h(t)$ is the spontaneous nuclear transformation rate of the radionuclide considered in the source region r_h .

$S(r_k \leftarrow r_h)$ is calculated by

$$S(r_k \leftarrow r_h) = \sum_i \Delta_i \cdot \Phi_i(r_k \leftarrow r_h) \quad (4.2.8)$$

where $\Delta_i = k \cdot n_i \cdot E_i$ is the mean energy emitted per unit cumulated activity, [J], with n_i the mean number of particles or photons per nuclear transformation, [(Bq·s) $^{-1}$], for radiation i , E_i the mean energy per particle or photon of radiation i , and k a numerical value for transformation of units; $\Phi_i(r_k \leftarrow r_h)$ is the specific absorbed fraction in target region r_k of radiation i emitted in source region r_h . The specific absorbed fraction $\Phi(v \leftarrow r)$ in a target volume v due to radiation emitted in a source region r is defined by

$$\Phi(v \leftarrow r) = \frac{\phi(v \leftarrow r)}{m_v} \quad (4.2.9)$$

where m_v is the mass of v , [kg], and $\phi(v \leftarrow r)$ is the absorbed fraction in a target volume v due to radiation emitted in the source region r .

In this formalism \tilde{A}_h is calculated with biokinetic models, see Sect. 4.2.4, and $S(r_k \leftarrow r_h)$ is calculated with dosimetric models, see Sect. 4.2.3.

4.2.2.2 ICRP formalism for public and occupational intake

Since its Publication 30 [79ICR] ICRP uses a similar method for dose assessment as MIRD which has been extended to age-dependence in its Publication 56 series [89ICR, 93ICR, 95ICR1, 95ICR2] but ICRP uses different notations.

The equivalent dose rate $\dot{H}_T(t, t_0)$ at age t in a target tissue T due to an acute intake at age t_0 is calculated by

$$\dot{H}_T(t, t_0) = \sum_S q_S(t, t_0) \cdot SEE(T \leftarrow S; t) \quad (4.2.10)$$

where $q_S(t, t_0)$ is the activity of the radionuclide considered in the source region S at age t after an intake at age t_0 , [Bq], which is calculated with biokinetic models, see Sect. 4.2.4, and $SEE(T \leftarrow S)$ is the specific effective energy which is the equivalent dose in the target tissue T per nuclear transformation in the source region S at age t , [Sv·(Bq·s) $^{-1}$].

The specific effective energy $SEE(T \leftarrow S; t)$ at age t is calculated by

$$SEE(T \leftarrow S) = \sum_R \frac{Y_R \cdot E_R \cdot w_R \cdot AF(T \leftarrow S; t)_R}{m_T(t)} \quad (4.2.11)$$

where Y_R is the yield of radiation R per nuclear transformation, $[(\text{Bq}\cdot\text{s})^{-1}]$, E_R is the energy of radiation R, [J], w_R is the radiation weighting factor, see Sect. 4.2.2.1, $AF(T \leftarrow S; t)_R$ is the absorbed fraction in the target tissue T per nuclear transformation in the source region S for radiation R at age t , which is calculated with dosimetric models, see Sect. 4.2.3, and $m_T(t)$ is the mass of the target tissue T at age t , [kg].

When equivalent dose rates at age t in a target tissue T are calculated the committed equivalent dose $H_T(70-t_0)$ due to an intake at age t_0 can be calculated by

$$H_T(70-t_0) = \int_{t_0}^{70} \dot{H}_T(t, t_0) dt \quad (4.2.12)$$

For adults the integration time to calculate the committed equivalent dose is 50 y, i.e. t_0 is taken to be 20 y. For non-adults an integration time until the age of 70 y years is considered. In nuclear medicine, however, the half-lives of the radionuclides involved are short, therefore in these cases the integration time may be infinite for all ages (see comment to eq. (4.2.3)).

4.2.2.3 Joint MIRD-ICRP formalism

The MIRD and the ICRP formalisms described above define the same dosimetry system even if it may not be well recognised by the reader because they use different terminologies. Therefore both groups defined a unified system which has been published as MIRD pamphlet No. 21 [09Bol] and which will be used by ICRP in their upcoming occupational intake reports which will replace ICRP Publications 30, 54, 68, and 78 [79ICR, 88ICR, 94ICR2, 97ICR]. It is also used here as the standard approach.

The committed absorbed dose in a target organ or tissue r_T is calculated by

$$D(r_T, T_D) = \sum_{r_S}^{T_D} A(r_S, t) \cdot S(r_T \leftarrow r_S, t) dt \quad (4.2.13)$$

where $A(r_S, t)$ is the activity in the source region r_S at time t and the S coefficient $S(r_T \leftarrow r_S, t)$ is the absorbed dose rate in r_T per unit activity in r_S . For adults the S coefficient is considered to be independent of time (age) and for short-lived radionuclides which are used in nuclear medicine its time-dependence can also be neglected.

The S coefficients are calculated by

$$S(r_T \leftarrow r_S, t) = \frac{1}{M(r_T, t)} \sum_i E_i \cdot Y_i \cdot \phi(r_T \leftarrow r_S, E_i, t) \quad (4.2.14)$$

where E_i is the (mean) energy of the i^{th} nuclear transition, Y_i is the number of the i^{th} nuclear transitions per nuclear transformation and $\phi(r_T \leftarrow r_S, E_i, t)$ is the absorbed fraction, the fraction of energy E_i emitted in the source region r_S at time t which is absorbed in the target organ or tissue r_T , and $M(r_T, t)$ is the mass of the target organ or tissue r_T at time t .

The quotient from the absorbed fraction and the target mass is called specific absorbed fraction:

$$\Phi(r_T \leftarrow r_S, E_i, t) = \frac{\phi(r_T \leftarrow r_S, E_i, t)}{M(r_T, t)} \quad (4.2.15)$$

The methods to calculate absorbed fractions are described in Sect. 4.2.3.

The absorbed fractions are also dependent on the type of radiation. For the calculation of committed equivalent doses radiation weighted S coefficients (S_w) are used instead of S coefficients. These are the S coefficients multiplied by the respective radiation weighting factors w_R :

$$S_w(r_T \leftarrow r_S, E_i, t) = \sum_R w_R \sum_i E_{R,i} \cdot Y_{R,i} \cdot \Phi(r_T \leftarrow r_S, E_{R,i}, t) \quad (4.2.16)$$

4.2.3 Dosimetric models

The aim of the dosimetric models is the determination of (specific) absorbed fractions for pairs of source regions r_S and target tissues r_T .

In general for non-penetrating radiation (α and - to less extent - β radiation) it is assumed that the absorbed fractions $\phi(r_T \leftarrow r_S) = 1$ for $r_T = r_S$ and $= 0$ for $r_T \neq r_S$. This is an adequate approximation for larger regions r_T and r_S . It is not adequate, for example, for small target regions as in the skeleton (see Sect. 4.2.3.2), the respiratory tract, the alimentary tract (see Sect. 4.2.3.3) and the bladder wall [10Zan].

For penetrating radiation (γ radiation) these values are calculated with Monte Carlo calculations which describe the photon transport within an anatomical phantom (see Sect. 4.2.3.1). For these calculations it has been assumed that the activity is homogeneously distributed within the source regions, and the average doses to the target tissues are calculated.

Now those calculations also are being performed for electrons and positrons with anatomical voxel phantoms [12ICR].

4.2.3.1 Anatomical models

Such anatomical phantoms have first been developed as mathematical phantoms which describe the body including its organs by geometrical figures like, for example, ellipsoids. The first mathematical phantom was the "MIRD phantom" of a hermaphrodite adult [69Sny, 78Sny], see Fig. 4.2.1

These phantoms later were extended to other ages resulting in a phantom family of the newborn, the 1, 5, 10, 15 year-old, and the adult [87Cri], see Fig. 4.2.2, which also shows cross-sections through the newborn and the adult.

These mathematical phantoms, of course, can only be rough approximations of the human anatomy but they were a big advantage compared to the former sphere calculations. With the development of computer power more realistic phantoms have been developed. These are Voxel (actually an abbreviation of volume pixel) phantoms which offer an improved anatomical representation of the human body compared to the MIRD-type mathematical phantoms. Voxel models require detailed anatomical data from patients of different age, sex, weight, height – data which can be obtained from CT or MRI examinations.

These phantoms describe an individual person. In radiation protection, however, doses to reference persons are needed as they are described in ICRP Publication 89 [02ICR]. For the male and female adult reference person ICRP has developed such voxel phantoms (see Fig. 4.2.3) on the basis of images of individuals who were similar in height and weight to the reference persons; the resulting voxel phantoms then were adjusted to the characteristics like organ weights of the reference persons. These phantoms were published as ICRP Publication 110 [09ICR]. The number of voxels is 211427 for the male phantom (voxel size 36.54 mm³) and 378204 for the female phantom (voxel size 15.25 mm³), respectively. In the near future ICRP will also develop voxel models for the non-adult reference persons.

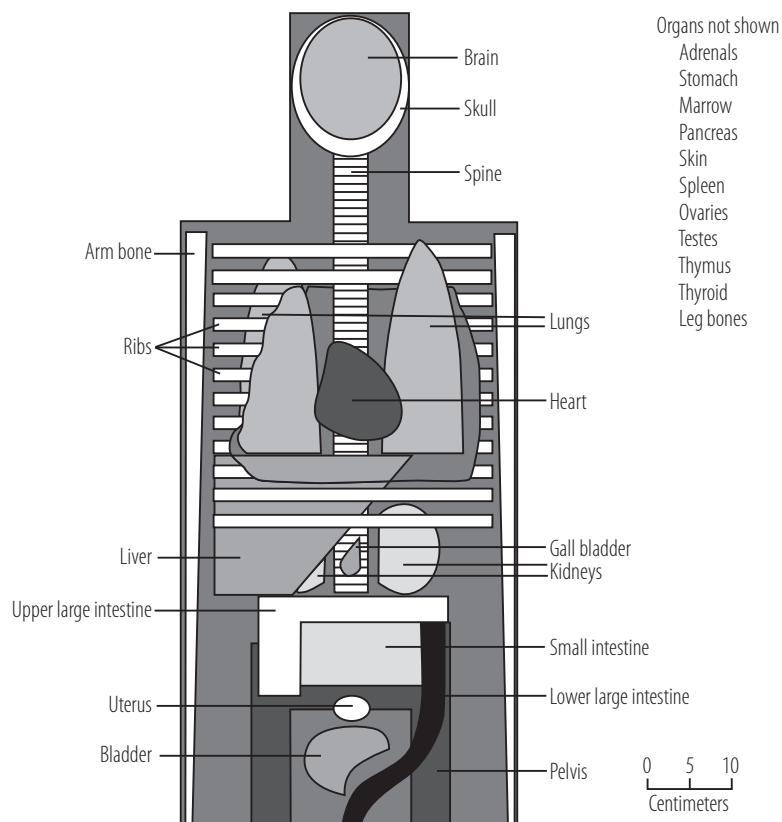


Fig. 4.2.1. Anterior view of the principal organs in the head and trunk of the MIRD phantom [78Sny].

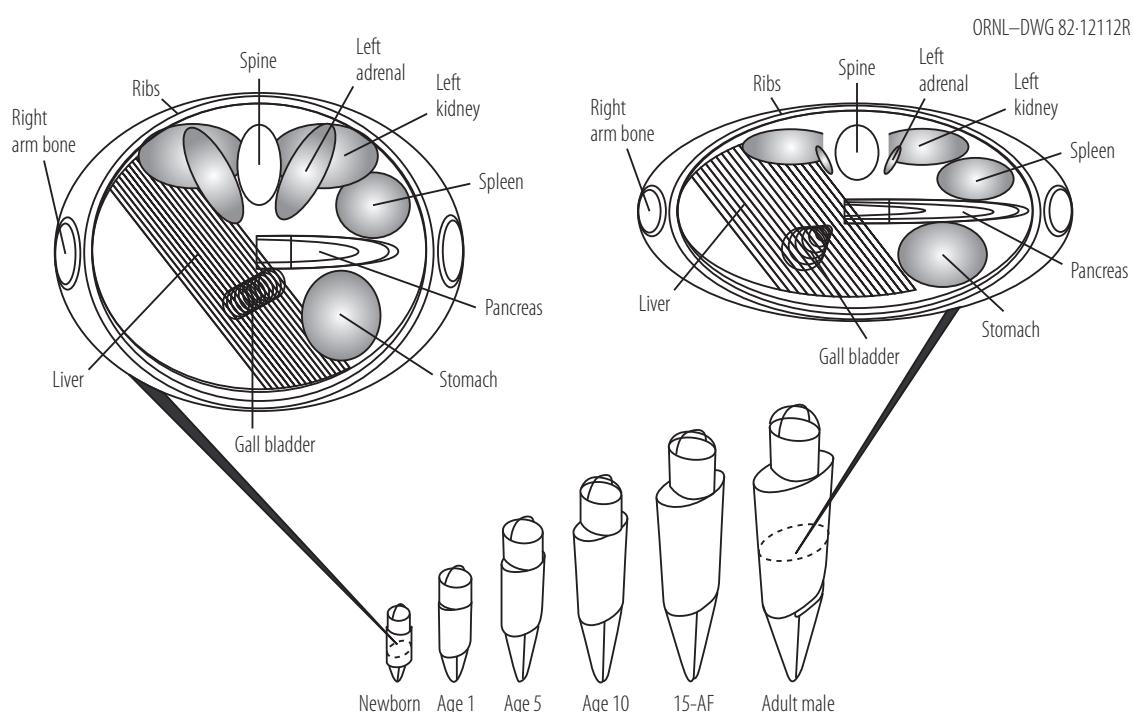


Fig. 4.2.2. External views of the phantoms and superimposed cross-sections within the middle trunk of the newborn and adult male phantoms [87Cni].

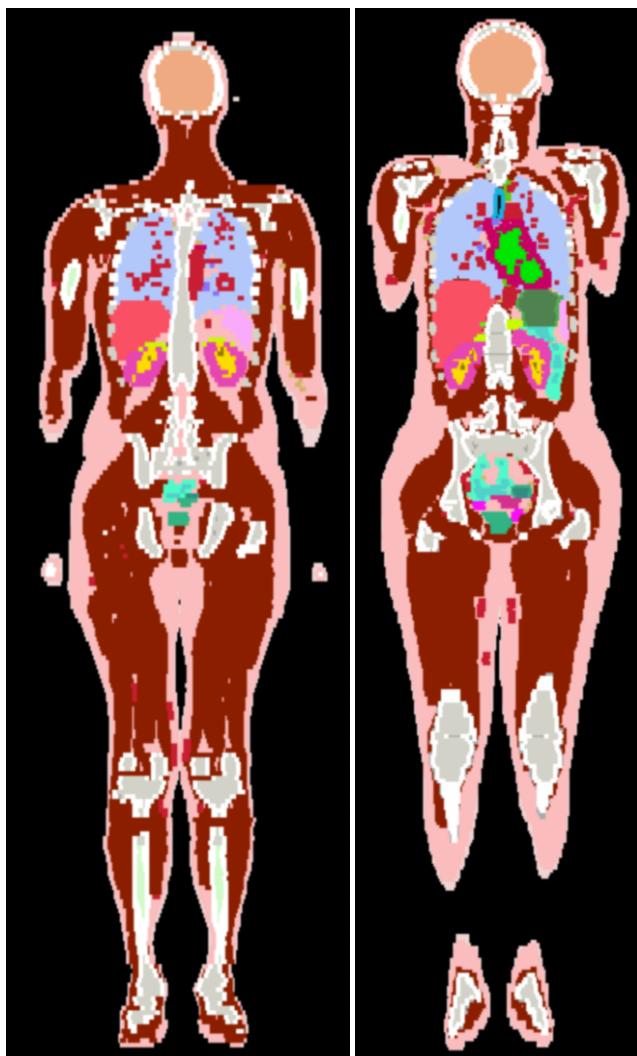


Fig. 4.2.3. Coronal image of the male (left) and female (right) reference adult [09ICR]

4.2.3.2 Models for bone

Until now, in bone, the source regions trabecular bone and cortical bone and the target tissues bone surfaces were considered. For the source regions it was distinguished between the assumption of the distribution of activity only on bone surfaces or within the whole bone volume. The target tissue bone surfaces were a $10\text{ }\mu\text{m}$ thin tissue layer at the surfaces.

In ICRP Publication 30 [79ICR] constant absorbed fractions for non-penetrating radiations (α and β radiation) were assumed, for β emitters on bone surfaces two values for low ($< 0.2\text{ MeV}$) and high mean energies.

At present the bone dosimetry is being much more refined. The new model averages the endosteal dose within $50\text{ }\mu\text{m}$ of the surfaces. For the calculation of absorbed fractions to endosteal tissues and red bone marrow micro-CT images of the various parts of the skeleton are used by coupling fluence-to-dose response functions with the particle fluence inside specific bone regions [12ICR].

4.2.3.3 Respiratory tract and alimentary tract

In the ICRP Human Respiratory Tract Model (HRTM) [94ICR1], see [Sect. 4.2.4.3](#), several target tissues within the extrathoracic and the thoracic part of the respiratory tract are considered. These are the anterior nasal passage (ET_1) and the posterior nasal passage, the pharynx and the larynx (ET_2) for the extrathoracic region and the trachea and bronchi (BB), the bronchioles (bb) as well as the alveolar interstitial region (AI) for the thoracic region of the respiratory tract. Additionally both, extrathoracic and thoracic lymph nodes are target tissues in the HRTM.

The target tissues are specified cell layers within the airways walls which are considered to be radiosensitive. These are basal cells of the epithelium in both extrathoracic regions, basal cells and secretory cells in the bronchial epithelium, Clara cells in the bronchiolar epithelium and endothelial cells, such as those of the capillary walls and type II epithelial cells, in the AI region.

ICRP Publication 66 [94ICR1] gives absorbed fraction values for non-penetrating (α and β) radiation for pairs of source regions and target tissues of the respiratory tract. For penetrating (γ) radiation until now the lung of the mathematical phantoms were taken for the derivation of AF values for all thoracic source regions and target tissues, and the thyroid as a surrogate for the extrathoracic source regions and target tissues.

In the development of voxel models [09ICR] the different sub-regions of the respiratory tract are considered and absorbed fraction values are calculated for γ radiation [12ICR]. For non-penetrating radiation the values of ICRP Publication 66 are still used because the target regions are too small to be modelled by voxels.

According to the methods described above regional doses to the respiratory tract are calculated. At the end doses to the extrathoracic region and to the thoracic region (lungs) are given. These are calculated as weighted mean values of the regional doses. The weighting is performed with so-called partitioning factors which indicate the radiosensitivity of the regional tissues. For the extrathoracic region the partitioning factors used are 0.998:0.001:0.001 for $ET_2:ET_1$:extrathoracic lymph nodes, and for the lungs the partitioning factors are 0.333:0.333:0.333:0.001 for BB:bb:AI:thoracic lymph nodes.

Similarly to the respiratory tract the new Human Alimentary Tract Model (HATM) [06ICR], see [Sect. 4.2.4.2](#), considers only specified radiosensitive cell layers as target regions. The location of the sensitive epithelial stem cells in the various regions is different for all regions of the tract and ranges from 60-100 μm for the stomach wall to 280-300 μm for the colon walls.

Absorbed fraction values for electrons are given in Annex F of ICRP Publication 100 [06ICR]. For α radiation the absorbed fractions are only different from 0 when source region and target tissue are the wall of the same site of the tract. In these cases the absorbed fraction is taken to be the fraction of the thickness of the layer of the radiosensitive cells and the thickness of the mucosa in the wall. For the small intestine (depth of the mucosa 200 μm , the radiosensitive cells are in a depth of 130-150 μm), for the example, the absorbed fraction would be $\phi(\text{small intestine wall} \leftarrow \text{small intestine wall}) = 0.1$.

The colon dose is calculated as the mass-weighted average of the doses calculated for the three colon segments right colon, left colon and rectosigmoid in the HATM.

4.2.4 Biokinetic models

For the calculation of the time-dependent activity content in source regions r_s , $A(r_s,t)$, biokinetic models are needed which describe the deposition and retention in source regions as well as the excretion pathways from the body. In general these are first-order compartment models described in [Sect. 4.2.4.1](#). There are biokinetic models recommended by ICRP for the alimentary tract, for the respiratory tract, for activity in the systemic circulation and those for the excretion pathways. These models are described in [Sects. 4.2.4.2 to 4.2.4.5](#).

4.2.4.1 General description of biokinetic models

In general, biokinetic models are formulated as compartment models, each compartment representing a source region which may be an organ or tissue or parts of them, or the contents of the respiratory tract, the alimentary tract, the gall bladder or the urinary bladder. A source region may also be represented by several compartments which are designated to different pathways or different retention times.

In the general case, the activity $A(i,t)$ in the i^{th} compartment of a chain of compartments at time t due to an intake at time $t=0$ can be calculated by

$$A(i,t) = \prod_{k=1}^{i-1} \lambda(k,k+1) \cdot \sum_{k=1}^i \frac{A(1,0) \cdot e^{-(\lambda_k + \lambda_R)t}}{\prod_{p=1, p \neq k}^i (\lambda_p - \lambda_k)} \quad (4.2.17)$$

where $\lambda(k,k+1)$ is the rate constant for transfer of material from compartment k to compartment $k+1$, $A(1,0)$ is the activity in the first compartment of the chain at time 0, λ_k is the rate constant for the description of total loss from compartment k to other compartments, and λ_R is the physical decay constant of the radionuclide considered.

Then the time-integrated activity $\tilde{A}(i,T_D)$ in compartment i can be calculated by

$$\tilde{A}(i,T_D) = \prod_{k=1}^{i-1} \lambda(k,k+1) \cdot \sum_{k=1}^i \frac{A(1,0) \cdot (1 - e^{-(\lambda_k + \lambda_R)T_D})}{(\lambda_k + \lambda_R) \cdot \prod_{p=1, p \neq k}^i (\lambda_p - \lambda_k)} \quad (4.2.18)$$

If there is a branching, then all branches can be considered separately. More complex biokinetic models, especially more realistic models considering physiological aspects, may be recycling models, i.e. that the movement may go back to any compartment which was involved earlier in the chain compartments. In those cases it may be sufficient to extend the chain of compartments by including a number of cycles of the recycling model which is dependent on the effective half-lives involved. The more exact method would be to solve the system of linear differential equations by the use of inverse matrices and / or by numerical methods.

The most precise way to formulate biokinetic models would be to give all the biokinetic transfer rates $\lambda(k,l)$ from a compartment k to a compartment l . More descriptive, however, is to present the biological half-time of a material in a compartment and to give all the branching fractions to other compartments to which the material is transferred. These both methods are equivalent: If the biological half time in a compartment k is T_b and the branching factor of movement to a compartment l is f_l then

$$\lambda(k,l) = \frac{f_l \cdot \ln(2)}{T_b} \quad (4.2.19)$$

On the other hand, if transfer rates $\lambda(k,l_i)$ are given from compartment k to compartments l_i with $i = 1, \dots, n$, then the biological half time T_b in compartment k and the branching fractions f_{l_i} are given by

$$T_b = \frac{\ln(2)}{\sum_{j=1}^n \lambda(k,l_j)} \quad (4.2.20)$$

and

$$f_{l_i} = \frac{\lambda(k,l_i)}{\sum_{j=1}^n \lambda(k,l_j)} \quad (4.2.21)$$

The loss of radioactivity from a compartment by transfer of the material to another compartment as well as by physical decay is described by the effective half-life T_{eff} which is defined by the biological half-time T_b and the physical half-life T_p :

$$T_{eff} = \frac{T_b \cdot T_p}{T_b + T_p} \quad (4.2.22)$$

4.2.4.2 Biokinetic models for the alimentary tract

At present, up to its Publication 106 [08ICR], ICRP uses the gastrointestinal tract model of ICRP Publication 30 [79ICR], see Fig. 4.2.4. This is a simple four compartment (stomach, small intestine, upper large intestine and lower large intestine) model based on Eve [66Eve] allowing absorption to body fluids (blood) from the small intestine compartment.

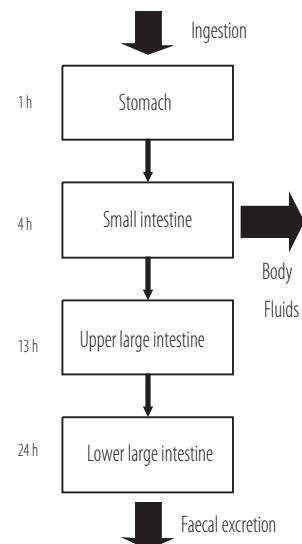


Fig. 4.2.4. Biokinetic model for the gastro-intestinal tract with mean transit times [79ICR].

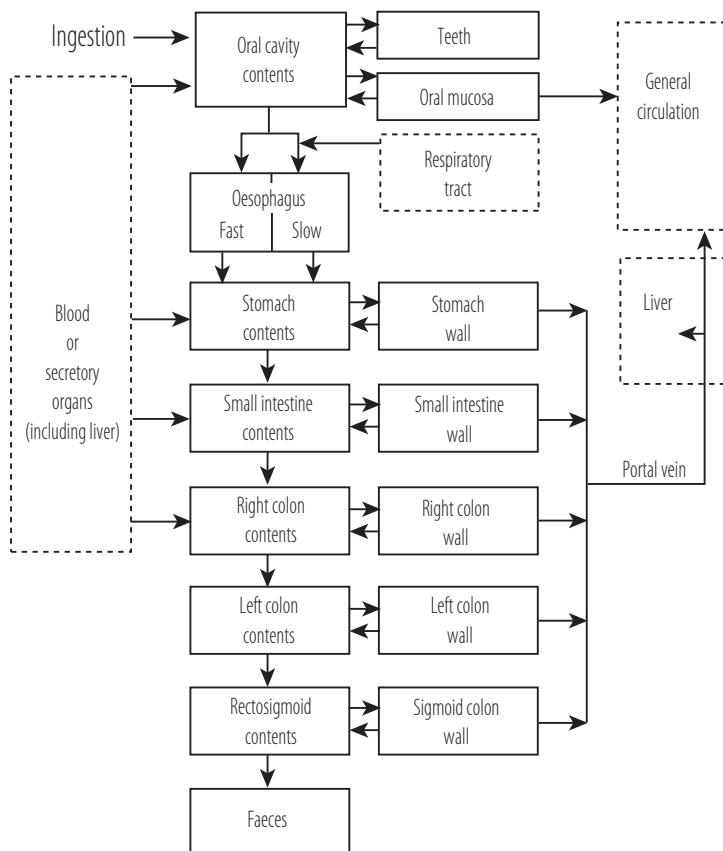


Fig. 4.2.5. The biokinetic human alimentary tract model by ICRP [06ICR]

In this model the mean transit times are 1 h for the stomach, 4 h for the small intestine, 13 h for the upper large intestine, and 24 h for the lower large intestine. These transit times are considered to be independent of the ingested material, and of age and gender of the person.

The fraction which is absorbed to body fluids from the small intestine is called f_1 and is dependent on the chemical properties (the solubility) of the material. In ICRP models for workers and members of the public it ranges from 10^{-5} for very insoluble material like insoluble compounds of plutonium to 0.99 for very soluble material like caesium- or iodine ions, which are nearly completely absorbed to blood.

In its Publication 100 [06ICR], ICRP has published a new Human Alimentary Tract Model (HATM), see Fig. 4.2.5, which will be used by ICRP from now onwards, first in the forthcoming Occupational Intake of Radionuclides (OIR) reports which will be published from 2012 onwards.

In contrast to the old gastro-intestinal model the new model also includes the oral cavity and the oesophagus which is important for the calculation of effective dose. Absorption to the systemic circulation is possible from (almost) all sites of the tract. This absorption process is not necessarily instantaneous as it was in the old model but there may be retention in the walls of the tract with subsequent recycling of parts of the material back into the contents of the tract. The total fraction of activity which is absorbed from the alimentary tract to the systemic circulation is called f_A in the new model which may be the sum of various local absorption fractions. If absorption takes place only from the small intestine without recycling back from the small intestine wall into the small intestine contents then f_A has the same value as f_1 in the old gastro-intestinal tract model.

The default mean transfer times of the HATM are shown in Table 4.2.2.

Table 4.2.2. Default mean transfer times in the HAT compartments according to ICRP Publication 100 [06ICR].

Age	3 months	1 year	5 – 15 years	Adult male	Adult female
Mouth					
– solids	-	15 s	15 s	15 s	15 s
– liquids	2 s	2 s	2 s	2 s	2 s
– total diet	2 s	12 s	12 s	12 s	12 s
Oesophagus					
– solids	-	8 s / 45 s	8 s / 45 s	8 s / 45 s	8 s / 45 s
– liquids	4 s / 30 s	5 s / 30 s	5 s / 30 s	5 s / 30 s	5 s / 30 s
– total diet	4 s / 30 s	7 s / 40 s	7 s / 40 s	7 s / 40 s	7 s / 40 s
Stomach					
– solids	-	75 min	75 min	75 min	105 min
– liquids caloric	75 min	45 min	45 min	45 min	60 min
– liquids non-caloric	10 min	30 min	30 min	30 min	30 min
– total diet	75 min	70 min	70 min	70 min	95 min
Small intestine	4 h	4 h	4 h	4 h	4 h
Right colon	8 h	10 h	11 h	12 h	16 h
Left colon	8 h	10 h	11 h	12 h	16 h
Rectosigmoid	12 h	12 h	12 h	12 h	16 h

For the oesophagus two mean transit times are given: for the fast component (90%) and for the slow component (10%) representing the residual activity after swallowing

4.2.4.3 The biokinetic model for the respiratory tract

The present ICRP Human Respiratory Tract Model (HRTM) is that published in ICRP Publication 66 [94ICR1], see Fig. 4.2.6. In this model the extrathoracic region and the thoracic region of the respiratory tract are considered. The extrathoracic region is sub-divided into the regions ET₁ (the anterior nasal passage) and ET₂ (the posterior nasal passage, the pharynx and the larynx), and the thoracic region (the lungs) are subdivided into the regions BB (the trachea and bronchi), bb (the bronchioles) as well as AI (the alveolar interstitial region).

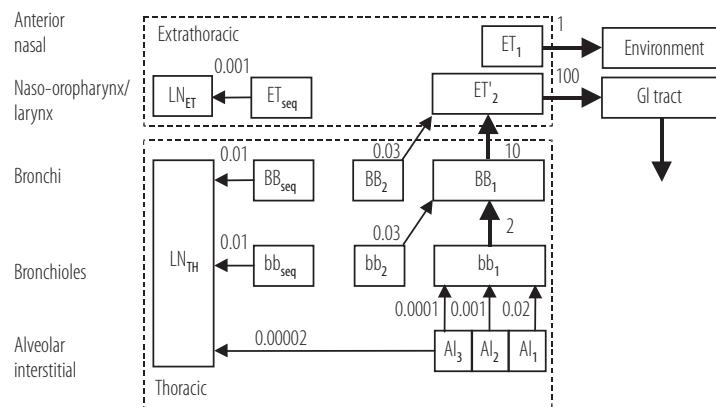


Fig. 4.2.6. Biokinetic Model of ICRP Publication 66 [94ICR1] for the human respiratory tract including mechanical transfer rates (in d^{-1}).

The deposition of aerosols depends on the physical properties of the aerosol, especially its size, and on the age, sex and breathing behaviour of the individual who inhales the aerosol.

In nuclear medicine often aerosols with particles of a fixed diameter are applied. In general, however, the particle sizes are variable and the size of an aerosol is given in terms of the AMAD, the activity median aerodynamic diameter. This means that half of the activity is related to particles with a higher aerodynamic diameter and the other half with particles with a lower aerodynamic diameter. The aerodynamic diameter of a particle is the diameter of a unit density sphere with the same terminal settling velocity in air as the particle considered. The AMAD is a suitable term for larger particles ($> 0.1 \mu\text{m}$) when the aerodynamic properties such as gravity and inertia are more relevant for the deposition processes. For smaller particles (up to $1 \mu\text{m}$) for which the thermodynamic properties of the particles such as the Brownian motion are dominating the deposition process the activity median thermodynamic diameter, AMTD, is more appropriate.

In cases of dose assessment for workers and general members of the public the AMAD is often not known. Then as a default assumption an AMAD of $1 \mu\text{m}$ is made for members of the public and of $5 \mu\text{m}$ for workers.

The methods to derive the deposition values from the physical aerosol properties like AMAD or AMTD, geometric deviation, density and shape factor are described in ICRP Publication 66 [94I1]. In Table 4.2.3 the default deposition values for workers and adult members of the public are listed.

Table 4.2.3. Reference deposition values for workers (AMAD $5 \mu\text{m}$) and adult members of the public (AMAD $1 \mu\text{m}$); these values are given for computational purposes to a higher precision as it would be justified by the underlying knowledge.

HRTM compartment	Worker	Adult member of the public
AI_1	1.596E-02	3.444E-02
AI_2	3.191E-02	6.888E-02
AI_3	5.319E-03	1.148E-02
bb_1	6.569E-03	9.879E-03
bb_2	4.384E-03	9.494E-03
bb_{seq}	7.721E-05	1.366E-04
BB_1	1.171E-02	6.796E-03
BB_2	5.921E-03	5.974E-03
BB_{seq}	1.243E-04	9.002E-05
ET'_2	3.989E-01	1.896E-01
ET_{seq}	1.996E-04	9.485E-05
ET_1	3.385E-01	1.489E-01

The fraction which is not deposited in any region of the respiratory tract is assumed to be expired again immediately.

For material deposited in the respiratory tract there are three clearance processes: Material deposited in the anterior nose (the ET₁ compartment) is removed into the air by extrinsic means (nose blowing, wiping, etc.). For all other material deposited in the respiratory tract there are two other competing clearance processes, mechanical clearance via the pharynx to the alimentary tract (input compartment is the slow compartment of the oesophagus) and via the lymphatic system to the lymph nodes, and absorption into the blood.

The mechanical clearance pathways and rates are shown in Fig. 4.2.6. A small fraction of material is retained in the airway walls, i.e. it is deposited in the sequestration compartments of the ET, BB and bb regions from where it is transferred to the extrathoracic or thoracic lymph nodes. A small fraction is also removed from the AI region (from the compartment AI₃) to the thoracic lymph nodes. Much more material is moved upwards by airway surfaces transport (mucociliary clearance) into the pharynx (included in the ET₂ compartment) where it is swallowed and transferred into the alimentary tract (into the slow clearance compartment of the oesophagus where it behaves due to the HATM). For this mechanical clearance to the pharynx fast and slow components are considered represented by different compartments. The distribution within the fast compartment, the slow compartment and the sequestration compartment is defined by the deposition model.

In the HRTM, the mechanical clearance is considered to be independent of the material and independent on the age and sex of the person considered.

The slowest mechanical clearance is in the AI region. The longest biological half-time is in the AI₃ compartment with about 5800 d or 15.8 y. However, as mentioned before, in each compartment except the ET₁ compartment there is the other competing clearance process, the absorption into blood.

The absorption to blood is considered to be in the same way from all compartments (except ET₁). Its structure is shown in Fig. 4.2.7.

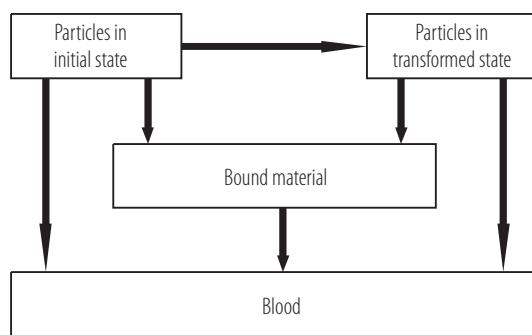


Fig. 4.2.7. Absorption processes from the respiratory tract compartments into blood [94ICR1].

From each compartment (here “Particles in initial state”) there is direct absorption to blood with transfer rate s_p . For a major part, however, there may be a slower absorption to blood. This part is transferred with transfer rate s_{pt} to the compartment “Particles in transformed state” from where it is absorbed to blood with the lower transfer rate s_t . Additionally the absorption model considers that the material may not be absorbed instantaneously into blood but may be retained within the walls. For this the compartment “Bound material” has been included: A fraction of the material is transferred into this compartment with the same transfer rates s_p and s_t and is absorbed into blood from there with the transfer rate s_b . In the compartment “Bound material” no mechanical clearance happens.

The absorption rates depend (only) on the chemical properties of the material. Because the material-specific absorption rates in many cases are not known, three default absorption types have been defined in ICRP Publication 60 [94ICR1] for compounds which are soluble, not so soluble, or insoluble. These default absorption types are the Types F, M, and S, because the absorption is fast, moderate, or slow. Sometimes additionally the absorption type V (very fast) is used. In this case an instantaneous absorption into blood is assumed. The absorption rates for the default absorption types F, M, and S are listed in Table 4.2.4.

Table 4.2.4. Default absorption rates of ICRP Publication 60 for absorption types F, M, S [94ICR1]

Absorption rates [d^{-1}]	Absorption types		
	F	M	S
s_p	100	10	0.1
s_{pt}	0	90	100
s_t	-	0.005	0.0001

While for soluble material (Type F) the material is absorbed very fast with a biological half time of 10 minutes, for insoluble material (Type S) most of the material is absorbed with a half-time of 19 y. However, it must be kept in mind again, that because of the competing mechanical clearance process the retention in the compartments is shorter.

In the default types the bound state is not considered. Until now it is also not used in the biokinetic data published by ICRP for various elements.

Inhaled gases and vapours are exhaled again when they are not dissolved in or in reaction with the airway surfaces. Therefore their deposition is dependent on their solubility and reactivity. There are three default classes defined in ICRP Publication 66 [94ICR1] for gases and vapours: Class SR-0 for insoluble and non-reactive material such as noble gases for which deposition in the respiratory tract is negligible, for these gases no deposition in the respiratory tract is assumed; Class SR-1 for soluble or reactive material like carbon monoxide or elemental iodine for which by default deposition values of 0.1, 0.2, 0.1, 0.2, and 0.4 are assumed for ET₁, ET₂, BB, bb, and AI, respectively; Class SR-2 for highly soluble or reactive material like carbon dioxide or tritiated water for which a complete deposition in ET₂ with immediate absorption to blood is assumed.

In ICRP Publication 53 [87ICR] biokinetic models are given for aerosols of $^{99\text{m}}\text{Tc}$, ^{111}In , and $^{113\text{m}}\text{In}$, as well as for the noble gases $^{81\text{m}}\text{Kr}$, ^{127}Xe , and ^{133}Xe . At that time the ICRP respiratory tract model [94ICR1] was not yet developed.

For the very short-lived $^{81\text{m}}\text{Kr}$ (half-life 13.1 s) there was the conservative assumption that all nuclear transformations of this gas would occur within the lungs. For the much more long-lived isotopes of xenon there are different models for a) single inhalation with 30 s breathhold, b) rebreathing for 5 min, and c) rebreathing for 10 min. A MIRD model [80Atk] considers retention of xenon in the body related to retention in lungs, lean body mass and two fat components.

For aerosols inhalation of particles smaller than 2–3 μm are considered in well-defined respiratory breathing patterns for which deposition mainly in the alveoli are assumed. There are models for soluble (like DTPA) and less soluble aerosols (like albumin). For these compounds biological half-times in the lungs of 1 h and 24 h are assumed, respectively.

4.2.4.4 Systemic models

Activity injected or absorbed into blood is distributed to various organs due to the chemical properties of the substance and then may be excreted. Fig. 4.2.8 shows the generic model of ICRP Publication 67 [93ICR] which is an extension of the generic model given in ICRP Publication 30 [79ICR] for systemic activity.

In this model activity is excreted directly from compartments representing the source organs instead of the physiologically more correct assumption that activity is transported back into blood and is excreted from there. Nevertheless this model type is appropriate for many cases especially for short-lived radionuclides and is frequently used for radiopharmaceuticals even in a more simplified structure: In radiopharmaceutical models often the blood retention and the delayed uptake into the source organs is neglected [87ICR]. In ICRP Publication 67 [93ICR] the excretion pathways were added. These are described in more detail in Sect. 4.2.4.5.

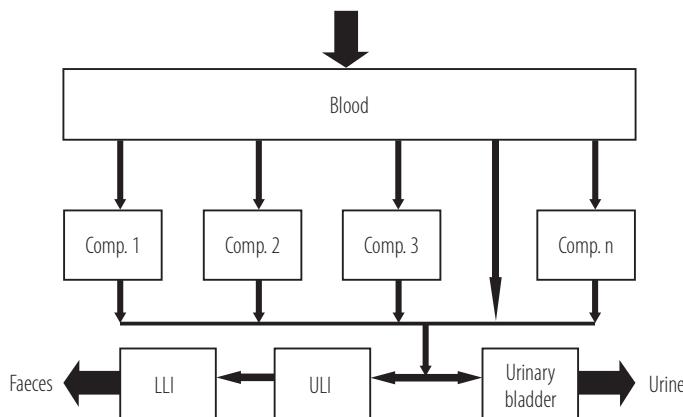


Fig. 4.2.8. Generic biokinetic model of ICRP Publication 67 [93ICR] for systemic activity; ULI and LLI are upper large intestine and lower large intestine, respectively.

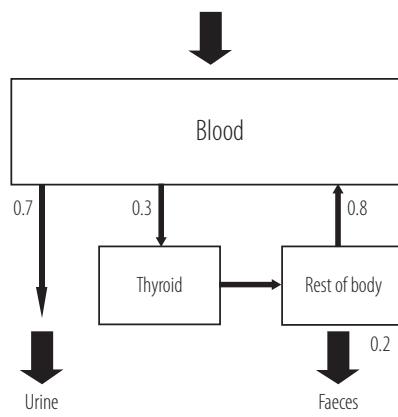


Fig. 4.2.9. Systemic biokinetic model for iodine [93ICR]

However, there are also more complicated models considering more the physiological behaviour of the substance. For iodine, for example, one of the first “recycling models” was developed [79ICR] which was supplemented by the excretion pathways later [93ICR], see Fig. 4.2.9.

In this model 30 per cent of iodine in blood is transferred to the thyroid whereas the rest is excreted in urine via the urinary bladder. Iodine is retained in the thyroid with a biological half-time of 80 d and is not excreted from there as it would be the case within the framework of the generic systemic model described above but it is moved as organic iodine to all tissues other than the thyroid where it is retained with a biological half-time of 12 d. 20 per cent of this organic iodine is excreted into faeces via the upper and lower large intestine while the remaining 80 per cent are recycled to blood. This iodine would again start to be taken up as inorganic iodine in the thyroid or excreted in urine.

The initial uptake of iodine in the thyroid is dependent on the health status of the person considered as well as on the content of stable iodine in the thyroid. In countries with iodine deficiency in food the iodine uptake would be larger, and the iodine uptake in the thyroid may be different for diseases of the thyroid. Therefore various thyroid uptake factors are considered in ICRP Publication 53 [87ICR] ranging from 0 % to 55 %. Recently, a more elaborated physiological systems model for iodine has been proposed for use in radiation protection [10Leg]. Another also more elaborated so-called recycling model developed on the basis of physiological knowledge is the iron model of ICRP Publication 69 [95ICR1], see Fig. 4.2.10.

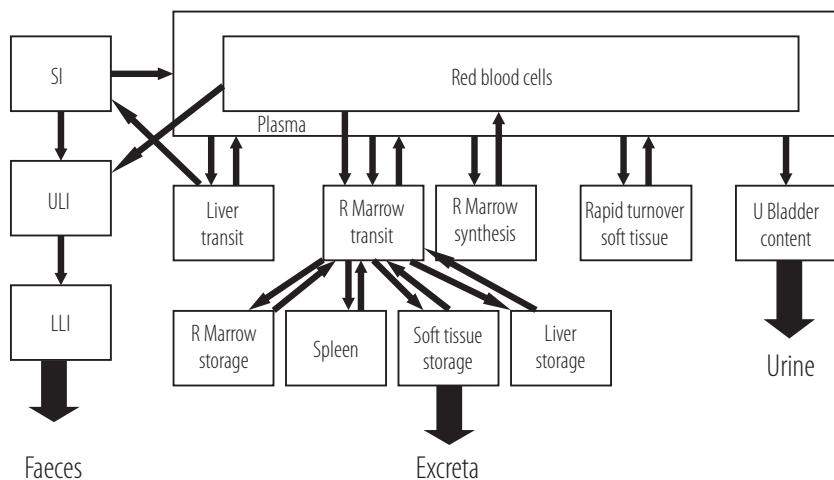


Fig. 4.2.10. Systemic biokinetic model for iron [95ICR1].

In this model the physiological behaviour of iron within the body is described: iron in blood is attached to red blood cells within the “Red Marrow synthesis” compartment. This iron is distributed in the reticuloendothelial system (red marrow, spleen, liver and other soft tissues) in their “storage” compartments via the “Red Marrow transit” compartment. After the destruction of the red blood cells this iron is recycled back to plasma via the “Red Marrow transit” compartment. The excretion pathway from the “Soft tissue storage” compartment is by exfoliation from skin.

Examples of biokinetic models for radiopharmaceuticals are given in Sect. 4.5.1.

4.2.4.5 Excretion models

The most common excretion pathways are via kidneys and urinary bladder to urine and via gall bladder and intestine compartments to faeces. Other excretion pathways – besides exfoliation from skin mentioned above in the iron model – are exhalation (especially for gases or substances metabolised to gases like ^{11}C and ^{14}C compounds metabolised to CO_2) or excretion via skin to sweat. Excretion pathways via nails and hair are only important for long-lived radioisotopes usually not used in nuclear medicine.

In the biokinetic models of ICRP for workers [94ICR2] and members of the public [93ICR] urinary excretion is generally considered via urinary bladder, and kidney retention is only considered in special models. For simplicity of the calculations the retention in the urinary bladder contents is described by a first order kinetics as it is done for all other compartments. The transfer rate from urinary bladder contents to urine is taken to be 12 d^{-1} for adults and older children which is in accordance with 6 voids per day or a bladder voiding interval of 4 h. For infants and 1 y old children this rate is taken to be 40 d^{-1} and 32 d^{-1} , assuming 20 and 16 voids per day, respectively.

In the ICRP models for radiopharmaceuticals described in ICRP Publications 53 and 106 [87ICR, 08ICR] a kidney-bladder model is used. For urinary excretion this model considers a mean transit time of 5 minutes through the kidneys with subsequent accumulation in the urinary bladder content. For this compartment no first order kinetics is considered but a realistic filling and voiding of the urinary bladder is considered. The voiding interval is taken to be 3.5 h for adults and children of age 10 y and 15 y. For 5-year-old children it is considered to be 3.0 h and for infants and 1-year-old children 2.0 h.

For faecal excretion an uptake of the excreted material is considered to be into the upper large intestine in the biokinetic ICRP models for workers [94ICR2] and members of the public [93ICR] which is then excreted into faeces via the lower large intestine according to the gastrointestinal tract model of ICRP Publication 30 [79ICR]. This is physiologically not very realistic (an uptake into the small intestine would be more realistic) but it was intended to avoid confusion with a possible re-absorption of material from the small intestine. Additionally, the dose to the small intestine to which no specified tissue weighting factor is attributed [91ICR] was considered to be of less importance.

In the ICRP models for radiopharmaceuticals described in ICRP Publications 53 and 106 [87ICR, 08ICR] an uptake into the small intestine was assumed with no re-absorption into the blood. Due to this assumption the material is excreted into faeces via upper and lower large intestine according to the gastrointestinal tract model of ICRP Publication 30 [79ICR]. For radiopharmaceuticals ICRP also considers biliary excretion into the small intestine via the gall bladder contents. For this pathway it is assumed that $\frac{3}{4}$ of the gall bladder content is cleared to the small intestine 3 h and 9 h after intake and all the content is cleared 24 h after intake.

4.2.5 Computer programs

There are various computer programs available for internal dosimetry, self-made programs as well as programs which are commercially available. For assessment of doses from radiopharmaceuticals the computer program OLINDA/EXM [05Sta] is by far the most common program. This program is succeeding the widely used MIRDOSE programs for which version 3.1 was the latest one [96Sta].

OLINDA/EXM can calculate doses for 814 radionuclides (including α emitters) for adult males and females (including pregnant females 3, 6, and 9 months after conception) and for children of five age groups from birth until 15 y.

There are different possibilities for the input of the kinetics: (a) the numbers of disintegrations in source regions per unit activity administered can be listed directly, (b) these values are calculated by the model when a biokinetic model is given (uptake and biological or effective half time in the source regions) or (c) these values are calculated with an extrapolation function when time-dependent measurement values of the activities in source regions are given.

For the calculation of the numbers of the disintegrations in the contents of the urinary bladder and the gastrointestinal tract some small sub-programs are included: For urinary excretion the body retention of the material excreted in urine can be given by fractions and biological half-times of body retention. Then the number of disintegrations in the urinary bladder content is calculated with a given assumption of the bladder voiding interval. For faecal excretion the input into stomach contents or small intestine contents can be given. Then the numbers of disintegrations in the contents of the gastrointestinal tract are calculated with the gastrointestinal tract model of ICRP Publication 30 [79ICR] with consideration of the fraction absorbed to blood from the small intestine if the intake to the tract was into the stomach content.

In the output of the program there are organ doses (in SI units and in old units rem and Ci) including the contribution of α , β , and γ radiation and their contribution to effective dose. The effective dose is given with the tissue weighting factors of ICRP Publication 26 [77ICR] (the "effective dose equivalent") and of ICRP Publication 60 [91ICR] (the "effective dose"). It is also possible to modify the doses by changing the target tissue masses.

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4.3 Collection of Biokinetic Data

Dosimetry in Nuclear Medicine Diagnosis and Therapy

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Assessing biokinetic data is essential in order to get a basis for estimating the absorbed dose from internally deposited radionuclides. Depending on the purpose for the dose calculation, diagnostic nuclear medicine, radionuclide therapy or occupational exposure, different strategies for the collection and assessment of biokinetic data should be applied. For diagnostic nuclear medicine, where individual doses are not to be calculated, data should be collected for a representative group of volunteers or patients. For the calculation of the internal dose the biokinetic data are usually to be presented in terms of uptake and retention in different organs. To obtain sufficient information normally repeated measurements are required, however, also data from single measurements may contribute with useful information, especially for physically short-lived radionuclides.

It is strongly recommended that absorbed doses presented in diagnostic nuclear medicine rely on biokinetic data collected in humans. However, in the case when introducing a new radiopharmaceutical, not yet tested on humans, one need to have some rough estimate of the dose before administering the substance to humans, and in this cases biokinetic data derived from animals may be used. Data thus obtained should be analysed in conjunction with a number of necessary assumptions concerning biological and physiological parameters, but also physical parameters, such as attenuation and scatter of photons, have to be considered.

The method that can be considered as adequate for measuring activity depends primarily on the type of radiation emitted, but also on the energy of the radiation, as well as the location of the radiation source. For measuring photon emitting radionuclides *in vivo* external probes or gamma cameras may be used, provided the presence of photons with proper energy (70 keV – 1 MeV), the upper limit depends on type of detector and collimator. In order to measure pure β -emitting radionuclides it is necessary to measure samples of blood, tissue or excreta, using e.g. a plastic scintillator or a liquid scintillation detector. In some situations the bremsstrahlung produced by pure β -emitters (e.g. ^{89}Sr for bone metastasis palliation) can be used for gamma camera imaging of the uptake.

4.3.1 Measurements with external probes

To obtain a complete picture of how the three-dimensional distribution of a radioactive substance in the body varies with time repeated site-specific measurements are required. Uptake and retention measurement *in vivo* using a non-imaging detection system is therefore of limited value, but since this is a simple and sometimes also a cheap method it may be useful in cases when the localisation of the source is well known. For measurement on humans, the most used and well-known application of a simple external detector is to assess the uptake of iodide in the thyroid. Another example is measurement of substances used for kidney clearance studies, where a detector located over one or both kidneys may be sufficient. It may also be a useful method to study *in vivo* whole body retention in small animals (mice). For a more accurate mapping of the activity distribution, this is, however, not an adequate method, e.g., for studying uptake and retention a radionuclide labelled antibody in a tumour for treatment purposes.

The most common detectors used for measuring activity *in vivo* are scintillation detectors, often based on a NaI(Tl) detector, but also other materials, such as CsI, or ZnS are being used. Also gas detectors, such as GM-tubes, may be used. Scintillation detectors allow the use of an energy window, this is an advantage, since pulses from background radiation and scattered radiation can be discriminated, which increases the accuracy of the measurements considerably. For low energy photons, typically 27 keV from ^{125}I , it is advantageous to use a thinner scintillation detector, since this reduces background registrations. Semiconductor detectors are even better in discriminating between direct radiation from the radionuclide

of interest and background, and may be used, but those detectors are more expensive and thus less available. In general the sensitivity of a detector increases with the increasing atomic number of the detector material.

Since we are interested in detecting activity located in a small area or volume, it is advisable to use a collimator mounted on the detector. A collimator will reduce the amount of background radiation, and radiation from other areas of the person measured that reaches the detector. The simplest collimator is merely a shield surrounding the detector allowing radiation to enter the front of the detector through an opening in the shield. Such an often used detector will have a comparatively large field of view. If a more limited field of view is desirable, a focusing collimator may be used, a shielded front with holes may be used, this will, however, decrease the sensitivity. The shield should be composed of a material possessing a high density and a high atomic number Z. Traditionally lead has been used, but lately also shield composed of tungsten has become popular. For low energy photons, copper or brass may be the material of choice. The thickness needed for an efficient collimator depends on photon energy, a collimator designed for ^{99m}Tc ($E_{\gamma} = 141 \text{ keV}$) is not sufficient for ^{131}I ($E_{\gamma} = 364 \text{ keV}$) or a PET-substance such as ^{18}F ($E_{\gamma} = 511 \text{ keV}$), which will need a considerably thicker material for an efficient collimation.

For a thorough review of different detectors the reader is referred to Chapter 10 in [05Kau].

4.3.2 Measurements of samples of blood or excreta

In addition to *in vivo* measurements blood samples may be collected with the purpose of assessing the retention of activity in blood. Additionally measuring the activity excreted in urine and faeces, is to assess the amount of substance leaving the system, and the rate at which it leaves. For these types of measurement scintillation or semiconductor detectors may be used. To enhance sensitivity the detector should be comparatively large and mounted shielded from background radiation with lead.

Since most substances are excreted mainly via the kidneys it is more common to collect and measure urine compared to faeces. It is also normally more readily collected and it is easier to get volunteers leaving urine samples. The total amount of activity excreted in one day should be assessed. It is important to be aware of that the urine production rate may influence the activity concentration in the sample. It is thus recommended that urine samples are collected as the total amount of urine produced during 24 hours.

See also Section 10.3.3 “In-vitro measurements” in [05Kau].

4.3.3 Whole body counting

For a radioactive substance which is distributed fairly homogeneously in the body whole body counting is the most sensitive method for measuring the activity in the body and its variation with time.

Whole body counting may be used as a valuable complement in establishing the total body retention, particularly in cases when one might suspect that there is a small long lived component. For short lived radionuclides, such as ^{99m}Tc or ^{123}I , with half-lives of 6.0 and 13.2 hours, respectively, this is, however, of limited interest. For substances labelled with more long-lived radionuclides such as ^{51}Cr or ^{75}Se it is important to assess the possible presence of a long-lived component of the biological retention function. E.g. for $^{51}\text{Cr-EDTA}$, which normally is rapidly excreted via the kidneys, the ICRP Publication 53 [87ICR] includes a fraction of 1 % of the administered activity that is retained with a biological half-time of 7 days, while the rest has a biological half-time of 1.67 hours.

For a more comprehensive description of the methods the reader is referred to Section 10.3.2 “In-vivo measurements” in [05Kau].

4.3.4 Gamma camera measurements on humans

The gamma camera is the preferred instrument for assessing biokinetics of radiopharmaceuticals in humans. Even if the spatial resolution is limited it is better than with most other methods available for *in vivo* measurement of the distribution of radiolabelled substances. This usefulness provides that photons with a proper energy are emitted from the radionuclide; this is not a problem for studying most radiopharmaceuticals, since they are designed for gamma camera imaging. Today planar imaging with gamma cameras is becoming less common than SPECT (single photon emission tomography) which is increasingly used. However, for quantitative measurements, when sometimes there is a need to image the whole body, time constraints and economy may be arguments for planar imaging instead of SPECT.

The activity required for gamma camera studies of biokinetic data on patients or normal volunteers, is similar to the amount normally used for diagnostic imaging, which means that the absorbed dose to the person studied, and hence the following risk, has to be considered when motivating these types of studies on volunteers. The use of a gamma camera for biokinetic studies of substances other than radiopharmaceuticals, which may involve more long-lived radionuclides, may be possible if the absorbed dose can be kept low enough. Together with supplementary measurements of blood and/or excreta, this is the most powerful method to achieve data on which a biokinetic model may be based.

Two main differences between imaging for biokinetics and normal nuclear medicine investigations may be noted:

- the “organ of interest” may be another compared to the diagnostic imaging,
- the need for repeated measurements in order to assess the retention at different points of time as a basis to derive functions describing the retention of the substance in different organs.

Additionally, in contrast to what usually is the case for nuclear medicine investigations, quantitative measurements are needed to assess the fractional uptake and retention in different organs. Such measurements may be based on phantom calibration measurements with known activity spatially distributed in a similar way as in the object (human or animal) studied. It is also important that the attenuation properties of the phantom are similar as in the object.

Another approach is to measure the syringe before the administration of the radiopharmaceutical. In that case the images have to be corrected for attenuation and scatter, using a validated method. A third alternative is to produce an initial whole body scan with the gamma camera immediately after injection, before any significant decay or excretion from the body. Such a scan provides information on the gamma camera response to the known administered activity.

The planar gamma camera images are usually not corrected for attenuation and the countrate registered by the camera will be dependent on the depth in the body at which the organ is located. The attenuation corrected count rate in an organ can be expressed as

$$C_{\text{att corr}} = C_{\text{net}} e^{\mu_{\text{eff}} d} \quad (4.3.1)$$

where μ_{eff} is the effective attenuation coefficient and d is the depth of the organ. The thickness of the organ is assumed to be infinitesimal. Values of the effective attenuation coefficient μ_{eff} are tabulated. For more specific applications, values of μ_{eff} can be directly estimated in the real experimental conditions by placing a planar source in a water phantom, varying the depth, plotting the count rate as a function of depth and fitting an exponential function.

A problem with gamma camera images, aside from the attenuation in the body, is how to subtract the background due to uptake in other organs and tissues properly. This problem arises mainly for planar imaging. Activity in circulating blood is also registered and is superimposed on the images; the correction for this is important in order to obtain data relevant for the uptake in the organ studied.

When tracer uptake in a small volume is measured, large biases can be introduced by the so called “partial-volume effect (PVE)” or what has recently been proposed to better be named “intensity diffusion” effect [09Skr]. The term describes the blurring of an ideal gamma camera, PET or SPECT activity distribution. The effects of smoothing with a 3D Gaussian filter during reconstruction of a PET image series may lead to a further diffusion of intensity: Choice of an increasingly wider smoothing filter makes the maximum signal from small tumours decrease (admittedly the noise decreases as well). With

good accuracy the resulting width of the point spread function may be calculated exactly. Mathematically, the reconstructed image intensity distribution can be described as a 3D convolution of the true sub-voxel activity distribution with the 3D gaussian point spread function. The term ‘intensity diffusion’ also provides an adequate description of 2D imaging with a scintillation camera equipped with a parallel hole collimator. The width of the diffusion (point spread) function depends on the distance from the source to the collimator. Again (for a point source in air) the maximal intensity (for equal duration of the acquisition) falls while the total counts remain the same.

4.3.4.1 Conjugate counting

Conjugate counting [79Fle, 99Sie, 10Gui] is the most common method for quantitative measurements using planar imaging, a method that often involves whole-body scans. For attenuation correction of planar imaging information is needed about the attenuating properties of the tissues. More specifically the depth to the organ/volume of interest needs to be known, together with the effective attenuation coefficient, which depends on the photon energy (e.g. 141 keV for ^{99m}Tc). The depth of the organ may be assessed by measuring from a perpendicular direction, or obtained from X-ray or CT images. A better way is, however, to eliminate the influence of this parameter, this is possible by imaging in opposite directions, usually from the front and from the back, using the geometric mean of the number of counts registered. This may require a two-headed camera in order to avoid uncertainties due to redistribution of activity between the imaging sessions. Today two headed cameras are common and it is easy to produce an image composed of pixels representing the geometric value for each pixel.

The count rate from the anterior view can be expressed as

$$C_A = \varepsilon A e^{-\mu_{\text{eff}} d} \quad (4.3.2)$$

where A is the activity in the organ and ε is the gamma camera system efficiency, d is the distance for the organ to the detector and μ_{eff} , which is energy dependent, is the effective attenuation constant for tissue.

The count rate in the posterior view can be expressed as

$$C_P = \varepsilon A e^{\mu_{\text{eff}}(T-d)} \quad (4.3.3)$$

where T is the thickness of the patient at the particular part of the body. In the conjugate view method the geometric mean of the anterior and posterior counting rate is used

$$\sqrt{C_A C_B} = \sqrt{\varepsilon^2 A^2 e^{-\mu_{\text{eff}}(d+T-d)}} = \varepsilon A e^{-\frac{\mu_{\text{eff}} T}{2}} \quad (4.3.4)$$

The organ depth is cancelled out and only the knowledge of the total thickness of the patient is needed. This equation can be rewritten so that the activity A in the organ is expressed as

$$A = \frac{\sqrt{C_A C_B}}{\varepsilon} e^{\frac{\mu_{\text{eff}} T}{2}} \quad (4.3.5)$$

4.3.4.2 SPECT and SPECT/CT

SPECT provides a method for *in vivo* quantification of the activity content that does not suffer the problems with projections overlapping with activity in background and in other organs. However, SPECT of the whole body is very time-consuming. Rotation around the whole patient for the collection of SPECT images may take around 40 minutes. Therefore, it is normally not practical to make whole-body SPECT images and SPECT measurements are presently normally limited to the most interesting sections of the body. One SPECT image taken over the organ(s) of highest interest for he absorbed dose estimation and used for activity quantification can be combined with the time activity curve from planar whole body images. For attenuation correction at ^{99m}Tc -imaging, earlier generations of SPECT scanners used sealed

line sources of ^{153}Gd ($T_{1/2} = 240$ d) or ^{57}Co ($T_{1/2} = 272$ d) to measure transmission through the body at detector elements on the opposite side of the body [11Mat]. These sources provide beams of around 100 keV photons and are scanned in the longitudinal direction at each step of the SPECT acquisition to provide transmission maps of the region under study. For ^{201}Tl imaging, ^{241}Am or ^{153}Gd sources are used in different configurations. Correction of images for attenuation effects are complicated by the broad range of tissue types (lung, soft tissue, muscle, and bone) that are present in the body volumes of interest. The goal of attenuation correction is to create a matrix of correction factors, each matrix element corresponding to a voxel in the SPECT image. The basis for calculation of such correction factors is an image set that contains in each voxel the value of the linear attenuation coefficient (a μ -map) for photons from the actual radiopharmaceutical (see below).

The clinical SPECT/CT systems currently used have typically dual head scintillation cameras and a CT sharing a common imaging table. There are two approaches to clinical SPECT/CT applications. The first is the use of a low-output fan-beam CT scanner, using a low current X-ray tube (2-3 mA). It can typically acquire up to four 5-mm anatomical slices in 10-15 s. The CT images acquired with this system are not of sufficient quality for diagnostic procedures, but adequate for attenuation corrections and anatomical correlation with emission images. The slow scan speed is actually an advantage in regions where there is physiological motion since the CT image blurring from the motion is comparable to that of the emission scans resulting in a good match in fused images. The effective dose from this system is typically around 1.5 mSv, which is much higher than for applications using radioisotope transmission sources (0.1 mSv).

The second approach is to integrate commercially available CT scanners with dual-head scintillation cameras [11Mat]. Typically today's scanners are with dual-head cameras, at present in one, two, six, 16 and 64 detector rows and with variable tube currents (20–500 mA), slice thicknesses of 0.6–12 mm, and rotational speeds of 0.5–1.5 s.

Since the CT scanners in these systems are commercially available diagnostic systems, the images produced are of sufficient quality to be used for diagnostic purposes, in addition to their obvious use for attenuation correction and anatomical correlation. Effective doses to the patient from examinations with these systems are in the order of 10–15 mSv, when images of diagnostic quality are produced. Operated in a lower radiation dose mode (about 3 mSv) by reducing the X-ray tube current these systems are still acceptable for attenuation correction and anatomical correlation applications. These images cannot be used for diagnostics, but are well suited for anatomical location of observed uptake besides for attenuation correction.

4.3.4.3 PET and PET/CT

Clinical PET/CT systems have CTs equipped with 4, 8, 16, 64 or even more detector rows and provide images of sufficient diagnostic quality to be used for diagnostic procedures. As with SPECT/CT systems, the CT scanners can be operated at reduced tube current if the scans are only to be used for attenuation correction and anatomical correlation.

For attenuation correction, earlier generations of PET scanners used rod sources of $^{68}\text{Ge}/^{68}\text{Ga}$ or ^{137}Cs that were rotated slowly around the patient to measure transmission through the body at detector elements on the opposite side of the body. Matrices of attenuation coefficients were then calculated by basically the same algorithms that are used in modern CT-scanners. Today, far more than 90% of the PET scanners that are purchased come with a CT scanner that is mounted on the same gantry as the PET scanner and the patient can be transferred under computer control between PET scanner and a helical CT-scanner [11Mat]. Again, corrections need to be carried out in order to take into account the differences in photon energies of 511 keV and the effective energy of the CT-scanner (around 70 – 90 keV).

Attenuation correction is simpler in PET than in SPECT because both annihilation photons need to reach detectors on the opposite sides of the body. The probability of this outcome depends on the sum of attenuation coefficients along the line that connects the two detectors where the two photons hit. This means that the acquired activity projections (sinograms) can be corrected for attenuation before the reconstruction.

4.3.4.4 Measurement, analysis and quality assurance of biokinetic data

The uptake, distribution and retention of the administered activity in the organs and tissues of the patients can be studied by acquiring gamma camera-, SPECT- or PET images over the body region of interest at different time points after the activity injection. It is common to collect blood samples as well as urine and sometimes faeces samples depending on the excretion of the radiopharmaceutical.

At which time after the administration of activity the images, blood, urine and faeces samples should be collected all depends on the radiopharmaceutical studied, its excretion rate and the physical half-life of the radionuclide [11Giu]. The time points for imaging depend on the effective half-life of the radiopharmaceutical studied. To study previously published biokinetic data of similar radiopharmaceuticals in humans or animals can work as a guide when setting up the imaging time scheme.

A general protocol could include one acquisition around the time of the effective half-life, and two additional acquisitions after around 3 and 5 effective half-lives. To make it possible to solve the compartment models (see next section) at least three data points are needed for each exponential terms of the model [11Giu]. It is important to include early time points in order not to overestimate the area under the curve. To characterise the long-term retention and not to underestimate the area under the curve late time points are necessary [11Giu].

It might be necessary to collect urine and faeces samples, depending on the excretion route of the radiopharmaceutical. It is more common to collect urine samples than faeces samples, partly because the kidneys are a common excretion route, but also because of the inconvenience for the patients to collect faeces. When collecting urine samples the cumulated activity is of interest. The patients can either be asked to collect each bladder voiding into different samples, divide the urine samples into different time intervals or collect all urine in one sample. The total duration of the urine collection depends on the effective half-life of the radiopharmaceutical in the body.

In planar gamma camera-, SPECT- and PET-images it is possible to draw regions of interest (ROIs) and obtain the count rate in a specific part of the image. In SPECT- and PET-images each organ is represented as a number of consecutive slices of known thickness. To determine the activity content in an organ one needs to draw one ROI in each of the slices containing the organ. The count rates in each slice are added together to obtain the total count rate in the organ. Such a procedure necessitates that the same reconstruction and filtering methods are used at measurements and at calibrations. Further there should not be any separation between the slices or any overlapping between them.

Neither measurement nor computations provide exact results. Therefore it is important to estimate the uncertainty of the biokinetic data. A consistent and generally agreed way to do it is to follow the Guide to the expression of uncertainty in measurement, GUM [95BIP, 06Sie].

4.3.5 Radiation protection, risk considerations when administering radionuclides to volunteers (non-patients)

To be able to do reliable dose estimates for patients, repeated quantitative measurements have to be done including measurements in addition to those done in the clinic. This means that the patient in this respect is a volunteer. To follow the retention and/or excretion of an administered radiopharmaceutical by additional measurement do not add any absorbed dose to the patient/volunteer. If however PET/CT or SPECT/CT technique is used, the extra CT-investigations will give a considerable addition to the absorbed dose. In rare situations, it is necessary to make investigations on “pure” volunteers, especially when new radiopharmaceuticals are going to be introduced.

Extra exposure of patients and exposure of volunteers is assumed to involve a small risk for radiation induced cancer (in addition to the risks from injections, blood sampling etc). To protect patients and individuals volunteering in such research from unacceptable risks, the World Medical Assembly has issued the Declaration of Helsinki, a set of ethical principles for the medical community regarding research involving humans, and is widely regarded as the cornerstone document of human research ethics. Based on this document, the radiological protection principles in biomedical research are discussed in detail in ICRP Publication 62 [91ICR]. The requirement of justification is considered in two steps: an

ethics committee approval of the research project and an informed consent of the individual volunteer. The ethics committee has to evaluate two specific questions: The overall merit of the research project to ensure that its design is appropriate for the task and that the scientific objectives are worthwhile and that the volunteer's radiation exposure is minimized, a question which need to be considered by a radiation protection committee and/or a qualified medical physicist. A cornerstone of the recommendations is the obtaining of informed consent from the prospective volunteer. The subject can withdraw this consent at any time.

Pregnant women should not be asked to take part in research projects involving irradiation of the foetus, unless the pregnancy itself is central to the research - and then only if other techniques involving less risk cannot be used instead.

4.3.6 References for 4.3

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4.4 Physical Properties of the Most Important Radionuclides

Dosimetry in Nuclear Medicine Diagnosis and Therapy

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Physical decay data given in this chapter are all taken from ICRP Publication 107 [08ICR]

For diagnostic nuclear medicine rather short-lived radionuclides with a suitable γ -emission are used. The by far most frequently used radionuclide is ^{99m}Tc with a half-life of 6.015 h and a main γ -emission of 141 keV (by an emission probability of 89.1%). Table 4.4.1 shows the half-life and the yield and energy of the most important radionuclides used in conventional diagnostic nuclear medicine.

Table 4.4.1. Physical properties of the most important radionuclides in conventional diagnostic nuclear medicine [08ICR].

Radionuclide	Half-life	Photon energy [keV]	Yield
^{67}Ga	3.26 d	93	0.392
		185	0.212
		300	0.168
^{81m}Kr	13.1 s	190	0.675
^{99m}Tc	6.02 h	141	0.891
^{111}In	2.80 d	171	0.907
		245	0.941
^{123}I	13.3 h	159	0.833
^{131}Xe	5.24 d	81	0.380
^{201}Tl	72.9 h	135	0.026
		167	0.100

A new diagnostic nuclear medicine procedure is the positron emission tomography (PET) which uses radionuclides which emit positrons. This positron emission is accompanied by the annihilation radiation which is characterised by two photons with energy of 511 keV, each in converse directions. These radionuclides in general are even more short-lived than the radionuclides used in conventional diagnostic nuclear medicine. Table 4.4.2 shows the half-life and the yield and energy of the most important positron emissions of the most important PET radionuclides.

Table 4.4.2. Physical properties of the most important PET radionuclides [08ICR].

Radionuclide	Half-life	Positron energy [keV]	Yield
^{11}C	20.4 min	386	0.998
^{13}N	9.97 min	492	0.998
^{15}O	122 s	735	0.999
^{18}F	110 min	250	0.967
^{68}Ga	67.6 min	836	0.877
		353	0.012
^{86}Y	14.7 h	535	0.119
		681	0.056
^{124}I	4.18 d	687	0.117
		975	0.108

For therapeutic applications of radiopharmaceuticals radionuclides with a significant non-penetrating radiation (mainly β radiation but also α radiation) are used. The most frequent therapeutic nuclear medicine application is thyroid treatment with ^{131}I . Other frequently used β emitters are ^{89}Sr , ^{153}Sm , ^{186}Re and ^{188}Re for bone palliation, as well as ^{90}Y and ^{177}Lu for the labelling of monoclonal antibodies which are accumulated in metastases. Their physical properties are listed in [Table 4.4.3](#).

Table 4.4.3. Physical properties of the most important β emitters used in radiotherapy [[08ICR](#)].

Radionuclide	Half-life	Mean β energy [keV]	Yield
^{131}I	8.02 d	182	1
^{89}Sr	50.5 d	585	1
^{90}Y	64.1 h	933	1
^{153}Sm	46.5 h	224	1
^{177}Lu	6.65 d	133	1
^{186}Re	3.72 d	347	0.95
^{188}Re	17.0 h	763	1

A few α emitters used in therapy are ^{223}Ra for bone palliation, ^{211}At for the labelling of monoclonal antibodies and ^{224}Ra for the treatment of Morbus Bechterew. Their physical properties are listed in [Table 4.4.4](#).

Table 4.4.4. Physical properties of some α emitters used in radiotherapy [[08ICR](#)]

Radionuclide	Half-life	α energy [MeV]	Yield
^{223}Ra	11.4 d	5.61	0.252
		5.72	0.516
^{224}Ra	3.66 d	5.68	0.949
^{211}At	7.2 h	7.45*), 5.87	0.576*), 0.417

*) Due to decay of the daughter nuclide ^{211}Po

Another α -emitter used in therapy outside nuclear medicine facilities is ^{222}Rn which is inhaled for the treatment of rheumatic diseases.

4.4.1 Reference for 4.4

08ICR ICRP: Nuclear Decay Data for Dosimetric Calculations. ICRP Publication 107 Oxford: Elsevier Science Ltd, 2008.

4.5 Doses to Patients in Diagnostics

Dosimetry in Nuclear Medicine Diagnosis and Therapy

D. NOBKE, S. MATTSSON, L. JOHANSSON

In diagnostic nuclear medicine it is neither meaningful nor recommendable to calculate the individual absorbed dose to the patient. Instead it is important to know the average dose to patients in various age groups for different investigations. This is a useful parameter for optimizing the radiation protection, e.g. with regard to which radiodiagnostic method to chose, and also for the authorities to calculate the collective dose from different diagnostic procedures.

4.5.1 Examples of biokinetic models for frequently used radiopharmaceuticals

For the purpose of internal dose calculations, biokinetic models, as well as anatomical and calculation models have been developed and presented by different authors [08ICR, 98ICR, 87ICR, 91Loe, 78Sny, 00Too]. These models can be used for calculating the cumulated activity in different source organs, and hence the absorbed dose utilizing methods described in Sect. 4.2. Since such models should be representative for the total population, the biokinetic models described are usually applicable for normal individuals. Even if it is not unusual with pathological changes of the biokinetics for nuclear medicine patients, these are normally not accounted for. However, the effect on the absorbed dose is often insignificant. In general, these models are not suitable for other purposes than dosimetry, e.g. to estimate the intake of a substance on the basis of internal and external measurements. It is also important to emphasize that they should not be used for dosimetry when the substance is administered for therapeutic purpose. In these cases parameters for the individual patients need to be determined.

For dosimetry the biokinetic data are usually presented in terms of fractional uptakes and biological half-times, i.e. first-order kinetics are assumed for the different organs. Thus, exponential functions are normally used to describe the retention function even if a compartment model in many cases would be physiologically more correct. In addition to the nuclide specific models a number of general models describing, e.g., the kinetics of the excretion routes (see Sect. 4.2.4) may be used. Many nuclear medicine products are excreted fast into the urine, and in those cases the assumption about the bladder voiding period may be a critical one. ICRP [87ICR, 98ICR, 07ICR] assumes a bladder voiding interval of 3.5 h for adults.

Biokinetic models designed for the purpose of dose calculations for radiopharmaceuticals are found in a number of publications. The MIRD-committee has produced and published biokinetic models for some of the most important radiopharmaceuticals in their “Dose Estimate Reports” (of which several are available on-line: <http://www.snm.org>). ICRP presents in their Publication 53 [87ICR], biokinetic models, as well as hence calculated doses, for most radiopharmaceuticals available at the time of printing, this report has thereafter been updated with recent substances in Publications 62, 80 and 106 [91ICR, 98ICR, 08ICR]. See Table 4.5.1.

4.5.2 List of dose coefficients

Dose coefficients [mSv/MBq] for a large number of nuclear medicine investigations are given by ICRP [08ICR, 98ICR, 87ICR] for adults and for 1, 5, 10, and 15 year-old children. Because in general there was no specific information available the same biokinetic data have been used for children as for adults with the exception of the bladder voiding interval. However, for children specific dosimetric data have been used obtained by calculations with (mathematical) children phantoms. Table 4.5.2 shows effective dose coefficients for adults for the radiopharmaceuticals listed in Table 4.5.1:

Table 4.5.1. Biokinetic models for some frequently used radiopharmaceuticals. For each radiopharmaceutical information about fractional distribution to a number of organs and tissues is given. Below that figure, the biological half-time for an uptake or elimination component is given. In parenthesis, the fraction in % of the fractional distribution taken up or eliminated with the corresponding half-time (in hours) is given (a minus sign indicates uptake).

Radiopharmaceutical	Investigation	Bone	Brain	Heart wall	Kidneys	Liver	Lungs	Spleen	Stomach wall	Thyroid	Other organs (Total body)	Excretion	Ref.
¹⁸ F-FDG 1.83 h	Tumour imaging, PET	8% $\infty(100)$	4% $\infty(100)$	5% $\infty(100)$	3% $\infty(100)$						80%	Urine	08ICR
¹¹¹ In-octreotide 67.3 h	Endocrine tumour imaging SPECT			6% 60(100)	6% 2.0(40)	5% 60(100)				0.1% 60(100)	82.9% 3.0(90)	Urine	08ICR
				60(30) 1680(30)	60(30)						$\infty(70)$		
¹²³ I-brain receptors 13.2 h	Imaging of neuro-receptors and transporters	6% 100(100)		3% 8(50)	20% 8(100)	20% 8(100)			5% 8(100)	0.3% 100(100)	45.7% 8(50)	GI-tract: 25% Urine: 75%	08ICR
⁵¹ Cr-EDTA 27.7 d	Renal function (normal)										100% 1.67(99) 168h(1)	Urine	08ICR
^{99m} Tc-DMSA 6.01 h	Renal imaging				50% 1(-100)	10% 1(-100)			1% 1(-100)		100% 2(25) 43(25)	Urine	08ICR
					∞ (100) 43(50)	2(50) 43(50)					∞ (50)		
^{99m} Tc-phosphates/ Phosphonates 6.01 h	Bone imaging	50% 0.25(-100)		2% 0.5(30)							100% 0.5(30)	08ICR	
^{99m} Tc-MAA 6.01 h		2(30) 72(70)		2(30) 72(40)							2(30) 72(40)	08ICR	
^{99m} Tc-albumin microspheres 6.01 h											100% 1.8(60)	Urine	08ICR
											36(40)		

Table 4.5.2. Effective dose coefficients for important radiopharmaceuticals.

Radiopharmaceutical	Effective dose coefficient [mSv/MBq]	Ref.
¹⁸ F-FDG	0.0019	08ICR
¹¹¹ In-octreotide	0.054	08ICR
¹²³ I brain receptors (generic model)	0.050	08ICR
⁵¹ Cr-EDTA	0.044*)	98ICR
^{99m} Tc-DMSA	0.0088	98ICR
^{99m} Tc-phosphates/phosphonates	0.0057	98ICR
^{99m} Tc-MAA	0.011	98ICR
^{99m} Tc-albumin microspheres	0.0061	98ICR

*) Oral administration

4.5.3 Diagnostic reference levels

The activity given to the patient in nuclear medicine diagnostics should be optimized considering radiation dose to the patient versus image quality. In practise the image quality, at least to some extent, depends on the number of registered counts. The consequence of a too small number of registrations is to get too noisy images, which will decrease detectability, and hence diagnostic quality. The number of counts is governed essentially by the activity administered and the time used for the investigation. Other factors of importance for the number of registered counts are radionuclide, time between administration and start of scanning, sensitivity of the gamma camera system – single- or two-headed system and type of collimator. Decreasing the activity administered, may be compensated for by a longer investigation time, or a more sensitive collimator. This may, however, result in a deteriorated image due to increased noise, and worse spatial resolution respectively. Furthermore SPECT and also ECG- and respiratory gated studies will need a larger number of registrations. To assess a proper activity to be given to the patient all these factors should be taken into account, including also practical considerations, such as the investigation time needed. Reference activities have been derived and established by different national professional bodies and radiation protection authorities (e.g. [[06ARS](#), [08Str](#), [04Nos](#)]). The reason for introducing such levels is to avoid the use of sub-optimized gamma-camera systems and methods for the image handling, since this may result in unnecessary exposure of patients. The reference levels should be considered as recommendations, if they are exceeded the user should present a good motivation for this, stating that this is necessary in order to obtain a satisfactory diagnostic quality. [Table 4.5.3](#) presents some national reference levels [[04Nos](#), [06ARS](#), [08Str](#)]. In nuclear medicine there is less work done concerning determinations and comparisons of image quality than in diagnostic radiology. There are few objective criteria defined of what should be seen in an acceptable image. The quality of nuclear medicine images is most often assessed through subjective judgements.

Table 4.5.3. Diagnostic reference levels (DRL) for adults for some frequently used investigations.

Investigation	Radio pharmaceutical	Sweden 2007 [08Str]	UK, ARSAC, 2006 [06ARS]	Germany 2003 [04Nos]
Thyroid imaging	^{99m} Tc-pertechnetate	120	80	75
Bone imaging	^{99m} Tc-MDP	600	600 800 (SPECT)	500 (benign) 700 (malign)
Myocardial perfusion (rest and stress)	^{99m} Tc MIBI/tetrofosmin	1200 2x600	800 1600(SPECT)	1000

Investigation	Radio pharmaceutical	Sweden 2007 [08Str]	UK, ARSAC, 2006 [06ARS]	Germany 2003[04Nos]
Myocardial perfusion (rest or stress)	^{99m}Tc MIBI/tetrofosmin	600	300 800 (SPECT)	600
Renal imaging	^{99m}Tc -MAG3	110	100	100
Renal imaging	^{99m}Tc -DTPA	150	300	150
Tumour imaging, PET	^{18}F -FDG	350	400	370 (2D) 200 (3D)

4.5.4 Paediatric patients

The absorbed doses to children are normally higher than to adults for the same administered activity [87ICR, 07ICR, 99Gad]. This is explained by the smaller mass of a child's body. On the other hand, due to the lack of adequate biokinetic data for children, most dose estimations are based on biokinetic data for adults [07ICR, 82Hen]. Since the metabolic rate for most substances are faster for children [02ICR], this approximation tends to overestimate the absorbed dose, but in total it will in general not completely compensate for the increased absorbed dose due to the smaller body mass. A child is also more sensitive to radiation compared to an adult, i.e. the probability of developing a cancer due to the exposure is larger, partly because of the higher proliferation rate, but mainly due to a longer expected remaining life length [07ICR]. Considering these facts it is understood that it is very important to optimise the activity administered to a child, balancing image quality and activity administered [91Pie]. In connection with this, it is also important to be aware of that due to the smaller mass less activity is needed for the same image quality when a child is investigated compared to what is needed for an adult [05Jac], the activity may thus be decreased without loss of diagnostic quality.

When estimating the activity that is to be administered to a child, it is important to have in mind that a too low activity could lead to unacceptable impaired image quality, this is also from a radiation protection point of view even worse than a slightly too high activity, since it leads to an exposure of no use for the patient. Various approaches have been taken in order to estimate the activity that should be given to child for a diagnostic nuclear medicine investigation. This calculation is today often based on the body surface area while other alternative parameters are the body mass and the age [90Pie, 98Smi, 05Jac]. The most recent recommendations from the EANM Dosimetry and paediatrics committee are now accompanied by a general rule of a minimum activity needed for acceptable or minimum standard on the image qualities [07Las, 08Las].

4.5.5 References for 4.5

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4.6 Doses to Patients in Therapy

Dosimetry in Nuclear Medicine Diagnosis and Therapy

D. NOBKE, S. MATTSSON, L. JOHANSSON

The most common application of radionuclide therapy is the treatment of hyperthyroidism by oral administration of ^{131}I . Cancer treatment with radioactive substances started already in the 1940-ies when treating thyroid cancer with ^{131}I . Since long, also ^{32}P has been used for treatment polycythaemia, and earlier also bone metastases. Today a number of radionuclides in different chemical forms are used for palliative treatment of skeletal metastases; the most common are $^{153}\text{Sm-EDTMP}$ (ethylenediaminetetramethyleneephosphonic acid) and ^{89}Sr chloride. Neuroendocrine tumours are treated with $^{131}\text{I-mIBG}$ (metaiodobenzylguanidine) or somatostatin labelled with ^{90}Y or ^{177}Lu . The introduction of antibodies for systemic therapy is slowly progressing, and today there are a few antibodies available, usually labelled with ^{131}I or the pure beta-emitter ^{90}Y , on the market mainly for lymphoma ($^{90}\text{Y}-\text{ibritumomab tiuxetan}$ (Zevalin[®]) and $^{131}\text{I}-\text{tositumomab}$ (Bexxar[®]) for refractory lymphoma) [06Mac, 10DeN].

4.6.1 Methods for dose estimation

In order to obtain tumour control in external radiation treatment the absorbed dose delivered to the tumour should be determined with a high degree of accuracy, if possible with an uncertainty less than 1-2 %. This high accuracy is, however, with presently available methods, normally not achievable in radionuclide therapy. This is partly due to shortages in the methods, but also to other reasons. The uptake of the test dose may differ from that of the treatment dose, not only due to effects caused by irradiation or higher activity concentration but also to normal biological variations.

For an optimal treatment with radionuclides an individual dose calculation needs to be performed in advance. Ideally, for this purpose an individual biokinetic study for the substance used is needed, primarily for other tissues but the tumour, especially for critical or risk organs [09Sjö, 09Dew, 08Sta]. The dose to the tumour is not always feasible to estimate, due to the spread of the disease. The result of such a study should then be used as the source for a Monte Carlo simulation of the absorbed dose, for this simulation a voxel phantom representing the individual patient can be used; such a phantom could be constructed from a whole body CT-study.

Another factor to have in mind when comparing absorbed doses in systemic radiotherapy is the fact that the dose calculated and presented is normally the average dose to the organ, and also the average to the cell. The exposure is however not completely homogeneously distributed, depending on the distribution of the radiation source. This problem has been reviewed by many authors, e.g. Thierens [01Thi].

4.6.2 Radioiodide therapy

The knowledge that the thyroid has a selective affinity for iodide was established in dogs in 1915 [49Ska]. The first studies of iodide metabolism in thyrotoxicosis were made in 1927 when stable iodide was administered orally and the iodine in the urine determined after different intervals. It was found that most patients with thyrotoxicosis excreted less iodine than normal subjects and this method (the iodine tolerance test) became frequently used before the production of radioactive iodide. In 1938, Hertz et al. [38Her] described the potential of radioactive isotope technique in the study of thyroid physiology and the investigations concerning thyroid diseases and iodide metabolism were enormously increased. Soon after ^{131}I became available from Oak Ridge in 1946 [95Ear], the first treatments were carried out by Chapman

et al. [46Cha] and Hertz et al. [46Her]. Early clinical investigations with radioiodine, initially using ^{128}I and later ^{131}I were made by Evans, Hamilton, Soley, Hertz, Roberts, Means and others [50Low, 56Mya, 95Ear]. Skanse introduced a method to determine the radioiodide uptake in the thyroid by analysing the 24-hour urinary excretion [48Ska].

There are 36 known isotopes of iodine, of which only one, ^{127}I , is stable. The first ones used clinically for diagnostic purposes were ^{128}I , ^{130}I , ^{131}I and ^{132}I [50Low, 56Mya, 95Ear]. They were produced either at cyclotrons ($^{130}\text{Te}+\text{D}$) or in nuclear reactors ($^{130}\text{Te}+\text{n}$) in Oak Ridge and Los Alamos, USA, and since 1947 also in Harwell, UK. Later ^{125}I was used for radioimmunoassay (RIA) and sometimes also for *in vivo* diagnostics as well as for therapy. Today the only isotope used for radiation therapy is ^{131}I , which together with ^{123}I also may be used in diagnostic nuclear medicine. ^{123}I has excellent physical properties for an imaging agent. It decays by electron capture with a half-life of 13 hours and emits photons with energy of 159 keV. It also provides lower absorbed dose to the thyroid with the necessary activity than does ^{131}I . With respect to the risk for stunning, it should therefore be an advantage to use ^{123}I instead of ^{131}I for imaging and uptake test purposes before a therapy. The major disadvantages of ^{123}I are high cost, and problems with availability and delivery from time to time. Despite these restrictions, ^{123}I should be the iodine of choice for thyroid imaging [02Par].

Additionally, ^{124}I is used for PET (positron emission tomography) studies, and possibly also other iodine β^+ emitters, e.g., ^{122}I and ^{121}I , may be utilised for this purpose in the future.

The normal way to quantify thyroid uptake of ^{131}I -iodide after *i.v.* or per-oral administration, has been to place a detector over the thyroid, where the activity very soon concentrates.

4.6.2.1 Calculations of the absorbed dose to the thyroid at treatment of hyperthyroidism

Radioiodine therapy of hyperthyroidism using ^{131}I -iodide is still the main form of therapy in nuclear medicine and is performed all over the world. For estimating the absorbed dose to the thyroid, information is needed not only about biokinetic parameters such as uptake and biological half-time, but also about the mass and geometrical shape of the uptake in thyroid tissue as well as decay characteristics for the radionuclide [06Mat]. For ^{131}I , the major fraction of the emitted energy, 67 %, is transferred to photons, while beta particles and electrons carry the remaining fraction. However, for a thyroid with a mass of 20 g, which is the mass of the thyroid of the “reference man” [02ICR], the major part of the absorbed dose is delivered by electrons and beta-particles. Photons contribute to approximately 5 % of the total dose, while for a mass of 100 g, which can be considered as a very large thyroid, the photon contribution has approximately doubled, but is still not responsible for more than about 10% of the total absorbed dose.

To obtain an accurate individual absorbed dose estimate for treatment of thyrotoxicosis, the active thyroid mass/volume should be assessed. This is often done by scintigraphic or ultrasound methods. For scintigraphy $^{99\text{m}}\text{Tc}$ -pertechnetate may be used, which has better imaging properties than ^{131}I and is cheaper and more readily available than ^{123}I , which, as earlier mentioned, should be a still better alternative. If available, SPECT imaging with a rotating gamma-camera is the best way to determine the mass/volume of the thyroid. Otherwise, the volume estimation has to be based on the projected area of the thyroid in an anterior image. Assuming that the thyroid is composed of two ellipsoidal lobes each with axes in the ratio of 2:1:1 the volume (V) can be calculated to approximately:

$$V = 0.38 a\sqrt{a} \quad (4.6.1)$$

with a density of 1 g cm^{-3} this equals the mass of the thyroid in grams, if the total projected area a is given in cm^2 . More specifically assigning axes in the ratio of $c:1:1$, the volume (mass) becomes:

$$V = \frac{4}{3\sqrt{2} c \pi} a\sqrt{a} \quad (4.6.2)$$

With the aid of dose factors for a uniform distribution of ^{131}I in small spheres, which can be obtained from the OLINDA internal dosimetry software (OLINDA/EXM uses the RADAR method of dose calculation

[05Sta] and the dose conversion factors as supplied on the RADAR web site (www.doseinfo-radar.com), the dose-rate \dot{D} in Gy/h from 1 MBq ^{131}I can be estimated to:

$$\dot{D} \approx \frac{0.117}{m} \quad (4.6.3)$$

where m is the mass in gram as calculated above. This is an approximation for small spheres. It should deviate less than 5 % from the real dose rate, and be good enough to be used also for the case when the activity is distributed within organs of other geometrical shapes, e.g. ellipsoids.

The total number of decays, the cumulated activity, can be calculated from the effective half-time, uptake and administered activity. Values for these parameters should be individually determined (see below). Putting it all together, using eqn (1) and (3) together with data on initial uptake and effective half-time, the absorbed dose D in Gy to the thyroid approximately becomes

$$\frac{D}{A_0} \approx 0.45 U T_{1/2} / a\sqrt{a} \quad (4.6.4)$$

where $T_{1/2}$ is the individual effective half-life in hours, U is the extrapolated initial uptake in the thyroid and A_0 is the administered activity expressed in MBq. As an approximation, it may be assumed that the uptake after 24 hours can be used. Taking it the other way around, the activity, A_0 , to be administered in order to achieve a desired absorbed dose is

$$A_0 \approx \frac{2.23 D a\sqrt{a}}{U T_{1/2}} \quad (4.6.5)$$

In the derivation of this equation a number of approximations has been done, which cause small inaccuracies compared to the usually rather large uncertainties in the individual determination of the thyroid mass, half-time and uptake of iodide. To improve the dosimetry, effort should be concentrated on the assessment of these parameters.

The three major parameters needed for absorbed dose calculations are, as discussed above; the volume of the functioning part of the thyroid, the uptake of ^{131}I -iodide in the thyroid and the biological half-time. All three parameters are individual and vary between patients, and they are all needed for an individual dose planning [05Jön].

When a protocol is used where the mean absorbed dose to the thyroid is not calculated for the individual patient, in general more activity than necessary is administered [05Jön]. This will unnecessarily expose the patient, the personnel and the patient's family. The excess patient exposure is of increasing concern since today younger patients are treated with ^{131}I , which might increase the risk of cancer during the rest of the patient's life. Also, since the absorbed dose to the thyroid is unknown, there is no parameter to correlate the therapeutic effect with.

To determine the exact absorbed dose to the thyroid, volume and biokinetics have to be taken into account. Concerning the uptake and half-time measurements, one can argue that the traditional repeated measurements (at 2, 4, 24, 48 h, and even more) are too time consuming. A simplified patient-specific protocol for treating hyperthyroidism has however been proposed [03Jön2]. It is based on single uptake measurement 4-7 days after the intake of a test activity of ^{131}I . This simplified protocol is as patient-convenient and time-effective as a protocol using a fixed administered activity. The methods for volume determination should be improved in general, since this parameter involves the largest uncertainties. Using recent SPECT/CT technology and ^{123}I , there are now improved methods to determine the functioning volume of the thyroid in a much better way than before. Volume estimation could be combined with a single uptake test – 4-7 days after the administration - using a small amount of ^{131}I added to the ^{123}I .

4.6.2.2 Dosimetry for treatment of thyroid cancer

The estimation of the absorbed dose to thyroid cancer remnants and metastases is still more complicated. One reason is the difficulty to define the distribution volume of the ^{131}I -uptake and its kinetics, due to possible spread metastases. Therefore fixed amounts of ^{131}I -iodide in the range of 2-7 GBq (typically 4 GBq) in a single administration are still normally used for the treatment of thyroid cancer. To be able to find possible correlations to the registered effects, the absorbed dose to thyroid cancer remnants and metastases should be calculated.

4.6.2.3 Thyroid stunning

Thyroid stunning means a temporary reduction in the ability of normal thyroid tissue or differentiated thyroid cancer to trap or retain a therapeutic activity of ^{131}I -iodide following a prior administration of a diagnostic activity of ^{131}I -iodide. The first observations related to thyroid stunning date back decades ago and several authors, who however have reported contradictory findings, have studied the phenomenon, mainly in connection with treatment of thyroid cancer, where comparatively high test activities are used. In an *in vitro* experiment using porcine thyroid follicle cells, Postgård et al. [02Pos] have shown that stunning is a real phenomenon, meaning a radiation induced and dose dependent reduction of the iodine transport capacity of viable thyroid cells (a nearly 50% reduction was observed already at 3 Gy). The data imply a potential reduction of the therapeutic efficacy of radioiodine treatment after diagnostic use of ^{131}I , even at low absorbed doses. Later work from the same group conclude that down-regulation of NIS is the likely explanation of ^{131}I -induced thyroid stunning. Enhanced NIS expression by synergistically acting agents (TSH and IGF-I) partly prevents the loss of iodide transport expected from a given absorbed dose, suggesting that thyroid stunning might be pharmacologically treatable [07Nor]. Therefore, it is important to further investigate the impact of thyroid stunning with respect to the absorbed dose for the various clinical situations, especially in the treatment and follow up of thyroid cancer [02Bre].

In a review article, Medvedic [05Med] concludes that thyroid stunning is evident in patients with well-differentiated thyroid cancer if thyroid remnants are irradiated by a few gray. The absorbed dose from the diagnostic administration of ^{131}I prior to radioiodine therapy should be limited to 4 Gy. Sabri et al. [00Sab] treated benign thyroid disease with two-step ^{131}I administration and proved that stunning really exists at the second one and that it depends only on the absorbed dose at the first ^{131}I application. The extent of stunning would be less than 1% (0.6%) for an absorbed dose of 35 Gy after the first administration, and thus negligible for practical purposes. These two examples, which show very different results, illustrate the need for further studies of the phenomenon. Possible differences in thyroid uptake of ^{131}I -iodide during the diagnostic test and the therapeutic situation could be detected by measuring the urinary excretion during the two procedures. In the therapy situation, it is easier to analyse the urine content of ^{131}I than to measure the high activity in the thyroid. There is obviously a need for more basic and clinically oriented studies regarding the stunning phenomenon, mainly in connection with treatment of thyroid cancer, but also in relation to treatment of thyrotoxicosis.

Since ^{123}I gives lower absorbed doses than ^{131}I , there is a growing body of evidence that this radionuclide is to be preferred for imaging, at least in connection with whole body screening in diagnosis of distant metastases from thyroid cancer.

4.6.2.4 Absorbed dose estimations to organs other than the thyroid for radiation protection purposes

For the estimation of the absorbed dose to different organs and of the effective dose, a biokinetic model is often used. Such models are designed to be representative for the general population for estimating the population averaged doses or the collective dose. Due to large individual variations, such models can *not* be used for individual dose estimations, neither to the thyroid, nor to other organs. In connection with

therapy, the absorbed dose to the thyroid has to be estimated based on individually assessed parameters, as discussed above.

Biokinetic models for iodine have been published by ICRP to be used both for occupational and environmental exposure [89ICR] and for nuclear medicine investigations [87ICR]. Other authors have also published models designed for different purposes. For dose estimation in nuclear medicine, the MIRD model by Berman et al. [72Ber] is an important contribution that has been widely used. ICRP [01ICR] has also published a model for iodine in the pregnant woman and the foetus.

In most models intended for nuclear medicine patients the uptakes in stomach wall and salivary glands are included. This is often not the case for the models intended for dosimetry of occupationally or environmentally exposed persons. On the other hand, models for patient dosimetry may be simplified in such a way that they are unsuitable for dosimetry of the long-lived isotope ^{129}I . Due to this discrepancy a more general compartment model, which can be used for different purposes, was recently developed [03Joh, 04Joh]. As mentioned above (Sect. 4.2.4.4), a more elaborated physiological systems model for iodine has recently been proposed for use in radiation protection [10Leg].

For treatment of thyreotoxicosis, there is still no consensus on an optimal protocol. Some patients are treated to obtain a normal thyroid function. Others are treated to obtain a function somewhat under the normal and some patients are made totally hypothyroid. The situation gets more complicated by the fact that the thyroid function gradually decreases with time after treatment and that hypofunction easily can be treated with thyroid hormone substitution.

Even without consensus regarding the desirable effect of the treatment, it is important both for the individual patient and for the understanding of the radioiodine therapy and its future optimisation that the absorbed dose to the thyroid is quantified [03Jön1]. The therapeutic effect of the therapy depends on the absorbed dose to the thyroid, just like in external radiation therapy. Comparing clinical observations without knowing the absorbed dose is not meaningful. If the outcome of a group of patients is to be studied, the result is futile unless the absorbed dose to the thyroid is known for each individual patient. For the individual patient it is also important not to be exposed to unnecessary radiation, but still to get the absorbed dose needed to the thyroid to avoid a second treatment.

4.6.3 Polycytemia

^{32}P is a pure beta-emitter with a relatively high energy on the beta-particles (max.: 1.71 MeV). It has been used since long for treatment of myeloproliferative disorders (as a myelosuppressive agent), and especially treatment of *polycytemia vera*, but also essential thrombocythaemia. Polycytemia is a chronic disease characterized by an increase in red blood cell mass. Today this substance is used for therapy mainly for elderly patients, due to the relatively high absorbed dose, especially the risk for acute leukaemia has been discussed [97Naj]. Phosphate with ^{32}P was the first radionuclide that was used for therapy, for treatment of leukaemia, in the late 1930-ies soon thereafter it was also used for treatment of Polycytemia [95Ear, 97Ber, 97Rob, 07Ten]. The phosphate ion is a bone-seeker; in the ICRP model a fraction of 0.3 of ^{32}P entering the blood is assumed to be taken up by the mineral bone where it thereafter is permanently retained [87ICR]. Together with a relatively high energy of the beta particles this makes the radionuclide suitable for radiotherapy of the bone marrow. ^{32}P is, however, also distributed into proliferating and protein-synthesizing cells, resulting in an enhanced absorbed dose besides in bone tissue also in liver and spleen [02Tho]. Usually the substance is administered orally, in the form of ortho-phosphate (PO_4^-) but it may also be given intravenously. The activity is generally given as a fixed activity per unit body surface with a maximum activity of 185 MBq; alternatively a fixed activity per body weight, with a maximum of 260 MBq may be used [07Ten].

4.6.4 Bone palliation

Bone pain arising from skeletal metastases from prostate or breast tumours may be treated with radiation from bone-seeking substances [05Fin, 08Bod, 10Pae]. Those are preferentially taken up in the metastases since the proliferation rate is higher here. Usually radionuclides used for this purpose are pure beta-emitters, such as ^{89}Sr , or they may emit a minor amount of gamma radiation, such as ^{153}Sm , which also allows visualizing the distribution of the substance [04Mai] using an ordinary gamma camera. Another beta emitting radionuclide that has been used for bone palliation is ^{186}Re [07Str]. Alpha emitters may also be advantageous, due to the short range and high energy of the alpha-particles, a bone seeking substance of special interest is therefore ^{223}Ra [07Nil, 06Bru], which may be used for this purpose.

The treatment is merely palliative, and the absorbed dose that may be obtained in the tumour tissues is limited by the dose to the red marrow.

4.6.5 Monoclonal antibodies and peptide and receptor specific substances

By labelling tumour specific monoclonal antibodies with radionuclides a potential tool for a powerful treatment, and hopefully an eradication of micro metastases and disseminated tumours, is obtained [07Bra]. These substances have been subject for research for a long period, and presently two commercially available labelled antibodies are ibritumomab tiuxetan labelled with ^{90}Y (Zevalin®) and tositumomab with ^{131}I (Bexxar®) [03Wis, 10Gol] for the treatment of non-Hodgkins lymphoma. For research, antibodies are usually labelled with radionuclides emitting beta-radiation, ^{90}Y , ^{131}I , ^{177}Lu etc., but also alpha emitters, such as ^{211}At , may show specifically advantageous properties for these purposes, since they will deliver a highly localized radiation dose.

Often today in clinical routine the dosage of the radioisotope in these cases are based on clinical studies of therapeutic effect and effect on normal tissues, like red bone marrow or kidneys. As a result of these studies a fixed activity per body weight has been established, that spares normal tissues from acute effects of the irradiation, but still has a therapeutic effect on the tumour. This is a simple but suboptimized way of dosing. In reality since a large margin is needed to the threshold for severe damage of the critical organ, the tumour dose may be too low to reach curative effect. An accurate dosimetry will therefore increase the probability for a successful treatment.

4.6.5.1 Neuroendocrine tumours

Neuroendocrine tumours may be treated with ^{131}I I-mIBG or somatostatin receptors labelled with ^{90}Y or ^{177}Lu [03Cap, 05For, 10Poo]. Treatment with ^{177}Lu labelled octreotate is a relatively new modality, which has increased significantly in frequency during the last years. For these treatments the kidneys are the critical organ. A review of dosimetric methods, especially kidney dosimetry including the assessment of residence time in the kidneys, has been published by Garkavij et al. [10Gar].

4.6.5.2 Treatment of liver tumours with microspheres

Selective internal radiation treatment (SIRT) is a treatment modality that is rapidly emerging [10Ahm]. Microspheres labelled with ^{90}Y are injected into the intrahepatic artery. The dosimetry is based on quantitative determinations of the tumour and liver volumes. A methodology for a more accurate dosimetry in this case has been described by Gulec et al. [10Gul].

4.6.6 Typical doses to tumours and critical organs

Typical absorbed doses to critical organs are presented in [Table 4.6.1](#). In cases where it is possible to give a typical dose to the tumour this is also displayed. In many cases, however, for disseminated tumours, with sometimes very small uptakes, this is not meaningful. Instead the therapy is given with the aim to maximize the dose to the tumour, which means that one has to limit the treatment solely due to exposure of critical organs. For the antibodies and peptides, repeated administration is usually needed in order to increase the probability for successful result.

Table 4.6.1. Typical activity and absorbed dose per administration for some common therapy agents (excluding iodide-131)

Substance	Typical adm. activity [MBq]	Tumor dose [Gy]	Critical organ 1 [Gy]	Critical organ 2 [Gy]	Ref.
⁹⁰ Y-Zevalin®	1000		Kidneys: 2.4	Red marrow: 2.7	09Fis
¹³¹ I-Bexxar®	3000		Thyroid: 8.1	Kidneys: 5.9	05Wah
¹⁵³ Sm- EDTMP	2500		Bone surfaces: 17	Red marrow: 3.8	08Bod, 87ICR
⁸⁹ Sr - chloride	150		Bone surfaces: 2.6	Red marrow: 1.7	08Bod, 93Ear
¹⁷⁷ Lu -octreotate	7400	200	Kidneys: 23		10Gar
³² P- phosphate	185		Red marrow: 2.0	Bone surfaces: 2.0	07Ten, 87ICR

4.6.7 References for 4.6

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4.7 Necessity of Patient-Specific Dose Planning in Radionuclide Therapy

Dosimetry in Nuclear Medicine Diagnosis and Therapy

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The basic goal of all radiation therapy is to deliver a lethal radiation dose to the treated tissue (most often a tumour) without causing undesired effects in normal tissues of the patient. In external radiotherapy patient-individualized dose calculations are an established routine. This is however not always the case when radiopharmaceuticals are used for therapy.

There are a number of reasons for individual dosimetry in radionuclide therapy:

- To get the same standard as in external radiation therapy and brachytherapy
- Patients have significant variability in their tumour and normal tissue uptake as well as in the rates at which the activity leaves these tissues
- Is there any uptake at all in the tumour?!
- To be able to correlate effects to absorbed dose
- Radiation protection (don't use more activity than needed)

Therefore, for all forms of radiotherapy with internal emitters, a patient-individualised dose calculation should be made for the most important tumours for which a specific uptake of the radiopharmaceutical can be made, and for the most important normal tissue at risk (normally the bone marrow, but possibly the lungs, kidneys, or other organs).

It is also important to note that individual patients' organ masses may differ significantly from those in the models of "reference" individuals available in standard dosimetry models, and that an estimate of the tumour and sensitive organ masses be made and adjusted for whenever possible.

After some period of time with such data gathering efforts, significant experience may be gained, with evaluation of the doses and the observed outcomes in a large number of patients.

This works fairly well in the use of radioiodines for thyroid cancer and hyperthyroidism, as the "therapeutic window" (difference in dose levels between what is experienced by the tumour and that experienced by the most important normal tissue) is large. It is a pity, however, that after decades of successful treatments with large numbers of patients, the use of a fixed activity approach has led to the fact that few investigators have characterized the radiation doses received by their subjects, so that an understanding of normal and diseased tissue response to radiation dose could be well established. In other forms of therapy, however (e.g., the use of monoclonal antibodies for radioimmunotherapy [RIT]), the tumour-to-normal tissue absorbed dose ratio may be low, and without the use of a patient-specific treatment planning strategy based on radiation absorbed dose, patients are mostly given low amounts of the therapeutic agent, to cautiously avoid deleterious effects in normal tissues (most notably the bone marrow).

The dosimetry of ^{90}Y -ibritumomab (Zevalin[®]) is a good example [07Cre, 09Fis]. For a more accurate calculation of the dose to tumour and also to critical organs, an individual dosimetry, based on measured activity distributions together with Monte Carlo simulation of the absorbed dose is needed [08He, 05Dew, 10Amr, 10Tso]. The more complex task is to map the time-dependent activity distribution, in three dimensions. For this reason a gamma- or positron emitting tracer may be given before the therapy, which should be studied using a proper imaging device, a gamma camera, or preferably, SPECT or PET. For ^{90}Y -ibritumomab, ^{111}In labelled ibritumomab may be used to predict the biokinetics of the ^{90}Y labelled substance [09Fis]. It is, however, important that the tracer used will follow the same biokinetics as the therapy substance and also that it does not affect the kinetics of the substance administered thereafter.

Data show that careful use of patient-individualized dose calculations will produce calculated radiation dose estimates that correlate well with observed effects and that use of a dosimetry-based approach will result in better patient outcomes, improving the quality of medical care for patients and reducing costs for the institutions involved [08Sta].

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4.8 Dose to Embryo and Foetuses in Diagnostic Nuclear Medicine Dosimetry in Nuclear Medicine Diagnosis and Therapy

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Nuclear medicine investigations are normally contraindicated during pregnancy. Radiopharmaceuticals are however occasionally administered to pregnant patients either due to clinical necessity or by mistake. In the first case the diagnostic test is of high importance for the maintaining of the health of the mother. In the second case an embryo or foetus may be irradiated unintentionally because the mother is not aware of her pregnancy, does not wish to admit it, or - against international recommendations [00ICR] - has not been asked whether she is pregnant.

The absorbed dose to the foetus or embryo will often be larger for a nuclear medicine investigation than for an alternative or corresponding X-ray or CT-study. An exception from this rule occurs when the X-ray or CT-study involves the lower part of the abdomen. In the nuclear medicine study the foetus often becomes exposed from radioactivity present in the excretion pathways. Most radiopharmaceuticals are excreted via the kidneys and urinary bladder, which are located close to the uterus [73Clo]. In the case of long-lived radionuclides and compounds excreted through the gastrointestinal tract, the problem might be even larger. However, most diagnostic nuclear medicine procedures make use of short-lived radionuclides (e.g. ^{99m}Tc) that do not cause large foetal doses.

Medical exposure of a pregnant patient has additional ethical considerations, compared to routine medical exposures. The situation for at least two individuals has to be considered in the risk/benefit analysis. A direct benefit for the mother has to be weighed against a possibly increased risk for the foetus, remembering that a benefit to the mother will also be beneficial for the foetus.

Thousands of pregnant patients are exposed to ionizing radiation due to different radiodiagnostic procedures, each year. Quite often this causes a great anxiety and sometimes unnecessary termination of pregnancy. For many patients, the radiodiagnostic examination and hence the exposure are justified, while for others the exposure may be inappropriate, since the unborn child may suffer from an increased risk. Exposure of patients who are pregnant is considered in detail by ICRP [00ICR, 08ICR].

The exposure of the embryo and foetus is dependent on (1) external irradiation from radioactivity in the mother's organs and tissues, (2) placental transfer of radiopharmaceuticals to the embryo or foetus and distribution of radiopharmaceuticals in embryonic or foetal tissues.

4.8.1 External irradiation from radioactivity in the mother

Radiopharmaceuticals retained by the mother – including possible uptake in placenta – and those which are not able to cross the placenta (e.g. radiocolloids) are external sources of irradiation to the foetus. Radiopharmaceuticals eliminated by the maternal kidneys are major sources of foetal irradiation since the urinary bladder is close to the foetus. In the absence of more specific information, the dose to all tissues of the embryo, up to the end of the second month of gestation, can be approximated by the absorbed dose calculated for the uterus.

Little information exists about the influence of pregnancy on the metabolism of radiopharmaceuticals compared to that of the non-pregnant female. Therefore no changes are normally assumed in the distribution and retention of radiopharmaceuticals in maternal organs during pregnancy, and the biokinetic models representing the average normal adult are used also for pregnancy. Absorbed dose estimates to the embryo-foetus include maternal contributions and foetal internal dose contributions from activity in the foetal placenta tissue.

4.8.2 Transfer over the placenta barrier

The physical, chemical, and biological properties of the radiopharmaceuticals are the critical factors in possible placental transfer. Radiopharmaceuticals (e.g. iodide, pertechnetate, iron, thallium and gallium ions) which are able to cross the placenta barrier and to concentrate in a specific organ could pose significant foetal risk [91Sta, 92Wat]. For all radiopharmaceuticals labelled with ^{99m}Tc and $^{123, 131}\text{I}$ there is also a risk to get free ^{99m}Tc -pertechnetate and $^{123, 131}\text{I}$ -iodide respectively, ions, which can easily cross the placenta barrier.

The transfer and exchange of radiopharmaceuticals between mother and the developing foetus has been reviewed by Stieve [83Sti, 85Sti, 87Sti]. These reviews have served as important basic material for the ICRP Publication 88 [02ICR], which mainly deals with substances, related to occupational exposure. Specifically for radiopharmaceuticals, Russell et al. [97Rus1, 97Rus2] have reviewed the literature and given information on placental cross-over of 15 radiopharmaceuticals from animal or human data. From these data, radiation dose estimates were developed in early pregnancy, and at 3, 6 and 9 months gestation for these radiopharmaceuticals and for many others used in nuclear medicine. For cases in which no information about placental transfer was available, the authors used data from animal studies to get – at least as a first approximation – information for estimations of doses to the human foetus or they provided dose estimates for foetal dose only from activity in the maternal organs.

In most publications only the average absorbed dose to the whole foetus is given. It is of course desirable to get a dose estimate to foetal organs, especially such as thyroid, liver and bone, where much higher dosed may occur. A significant uptake of radioiodide by the foetal thyroid will occur after the 10th week after conception [01ICR]. The uptake within the thyroid increases with the activity of the gland; from 0.025% during the 3rd month to about 2% at term [87Sti].

4.8.3 Phantoms for pregnant women

Anthropomorphic models representing the pregnant woman at 3, 6 and 9 months were developed by Cristy and Eckerman [87Cri, 95Sta]. Several other groups have further extended the modelling approach and also calculated specific adsorbed fractions (SAF) for these models. The work of these groups are referred to in a recent paper, where Shi et al. [08Shi] used newly developed models representing pregnant females at 3, 6, and 9 month gestational periods to derive the SAF values for internal photon emitters. This set of deformable models allowed the organs to be easily adjusted to match with the ICRP reference organ masses recommended for an average pregnant female and her foetuses [89ICR].

4.8.4 Dose coefficients

Dose coefficients [mGy/MBq] for selected radiopharmaceuticals have been published [95Mat, 97Rus1, 97Rus2, 87ICR, 98ICR, 08ICR]. See Table 4.8.1.

Table 4.8.1. Typical mean whole body foetal doses for some common radiopharmaceuticals in early pregnancy and at term. Dose includes maternal and foetal internal dose contributions. (From IAEA http://rpop.iaea.org/RPOP/RPoP/Content/SpecialGroups/1_PregnantWomen/PregnancyNuclearMedicine.htm).

Radio-nuclide	Procedure (substance)	Administered activity [MBq]	Early pregnancy [mGy]	Nine months [mGy]
^{99m} Tc	Bone scan (phosphate)	750	4.6-4.7	1.8
^{99m} Tc	Lung perfusion (MAA)	200	0.4-0.6	0.8
^{99m} Tc	Lung ventilation (aerosol)	40	0.1-0.3	0.1
^{99m} Tc	Thyroid scan (pertechnetate)	400	3.2-4.4	3.7
^{99m} Tc	Red blood cell	930	3.6-6.0	2.5
^{99m} Tc	Liver scan (colloid)	300	0.5-0.6	1.1
^{99m} Tc	Renal function examination (DTPA)	750	5.9-9.0	3.5
⁶⁷ Ga	Abscess/tumour imaging (citrate)	190	14-18	25
¹²³ I	Thyroid uptake ¹⁾ (iodide)	30	0.4-0.6	0.3
¹³¹ I	Thyroid uptake ¹⁾ (iodide)	0.55	0.03-0.04	0.15
¹³¹ I	Metastases imaging ¹⁾ (iodide)	40	2.0-2.9	11.0

¹⁾ Foetal thyroid doses are much higher than foetal whole body dose, viz. 5-15 mGy/MBq for ¹²³I and 0.5-1.1 Gy/MBq for ¹³¹I.

4.8.4.1 ¹⁸F-FDG

In recent years PET and PET/CT has become an essential component of cancer management. Millions of ¹⁸F-FDG PET examinations are performed yearly worldwide and in rare cases of unintended exposure of such examinations during pregnancy. ¹⁸F-FDG data for maternal and foetal uptake in primates from Benveniste et al. [03Ben] were used to estimate human foetal doses to 0.022 mGy/MBq in early pregnancy and at 3 months gestation and 0.017 mGy/MBq at 6 and 9 months. More recent case reports based on careful kinetic and dosimetric information give 0.033 mGy/MBq to an embryo [08Zan] and 0.04 mGy/MBq in early pregnancy [10Zan]. These values are somewhat higher than those reported from MIRD [02Hay]: 0.018-0.022 mGy/MBq for early pregnancy.

For some radiopharmaceuticals like ^{99m}Tc-pertechnetate, -DMSA, -DTPA, -HDP, -MAA and -MAG₃ there are model-based comparisons of maternal and foetal organ doses [02Sau]. Data for ^{99m}Tc pertechnetate is given as an example below.

4.8.4.2 ^{99m}Tc sodium pertechnetate

The relation between mean absorbed dose in foetal organs and activity given to the mother is given in Table 4.8.2 [02Sau].

Table 4.8.2. Estimated mean absorbed dose per unit activity given to the mother for maternal and foetal organs following intake of ^{99m}Tc -sodium pertechnetate in the first and third stage of pregnancy. (From [02Sau]).

Organ	Absorbed dose [$\mu\text{Gy MBq}^{-1}$]		
	Maternal		Foetal ^{a)}
	Stage 1	Stage 3	
Adrenals	4.64	4.19	5.03
Brain	1.25	1.09	2.53
Gallbladder	4.86	4.54	5.30
L L int.	5.34	3.8	4.97
Small int.	4.82	4.16	5.15
Stomach	4.01	3.78	5.00
U L int.	4.42	3.00	5.10
Heart	3.29	3.07	4.45
Kidney	4.43	3.74	4.96
Liver	3.84	3.75	5.78
Lungs	3.01	2.70	5.05
Muscle	3.78	3.24	4.60
Red marrow	3.94	3.56	4.30
Bone surface	6.82	6.14	5.84
Skin	2.60	2.31	3.97
Spleen	2.85	2.80	5.97
Thymus	4.54	3.07	5.60
Thyroid	9.18	6.27	139
Urinary bladder	22.8	28.4	4.85
Total body	3.98	3.67	4.54
Uterus	5.78	2.61	5.14
Placenta	5.78	3.88	
Foetus	5.2 ^{b)}	4 ^{a,c)}	4 ^{a,d)}

^{a)} 42.5% maternal contribution of $1.9 \mu\text{Gy MBq}^{-1}$

^{b)} Includes 77.1% maternal contribution of $4.04 \mu\text{Gy MBq}^{-1}$

^{c)} Uses whole foetus organ

^{d)} Built up from foetal organs

4.8.5 Recommendations to decrease dose to foetus

The use of smaller administered activities and longer imaging times can reduce the absorbed dose to the foetus. This is feasible if the patient is not too sick and is able to remain resting.

Since radionuclides in maternal tissues contribute to foetal dose, maternal hydration and frequent voiding can reduce the foetal dose after the administration of a number of radiopharmaceuticals.

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4.9 Doses to Infants From Breastfeeding

Dosimetry in Nuclear Medicine Diagnosis and Therapy

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Breast milk is the ideal nutrient for the newborn, but unfortunately also a route of excretion for some unwanted substances, like pharmaceuticals (including radiopharmaceuticals), alcohol, heavy metals and volatile organic compounds, as well as persistent, bioaccumulative, and toxic chemicals.

Generally, administration of radiopharmaceuticals to a breastfeeding mother is avoided as activity in the breast milk causes an unwanted radiation exposure of the infant who ingests the milk. This is of special concern since infants have a comparatively high sensitivity for radiation and not the same direct benefit of the investigation as the mother. When it is vital for the mother to have the examination performed it is, in some situations, necessary to interrupt the breastfeeding temporarily. This interruption may cause feeding problems and beyond a certain length of time, it will be difficult or impossible for the mother to maintain her milk supply and thus to be able to take up breastfeeding again. Unfortunately, too often, the breastfeeding is terminated even though it is not necessary. On the contrary there are examples of situations when a necessary termination has been ignored. It is therefore important to have access to proper and clear recommendations on the duration of a feasible breastfeeding interruption. Apart from dietary information such recommendations should be based on accurate biokinetic and dosimetric data.

4.9.1 Excretion

Systematically collected data on the excretion of radionuclides into breast milk are rare. Data are often published in the form of case reports. Taken together, there are now data for the most frequently used 99m Tc-labelled substances and it has also been possible to do estimates for a number of other substances [60Wea, 73Wyb, 81Mat, 85Ahl, 90Ros, 94Rub, 10Lei].

4.9.2 Estimated doses

To be able to make a proper estimate of the absorbed dose to the child it is important to know the chemical form of the radionuclide in the breast milk. This is seldom known or analysed and therefore more or less conservative assumptions have to be done. For 99m Tc-labelled radiopharmaceuticals, there are reasons to assume that a majority of 99m Tc in breast milk is in the form of 99m Tc-pertechnetate and for $^{123},^{131}$ I-labelled radiopharmaceuticals in the form of $^{123},^{131}$ I-iodide [10Lei].

When available, radiopharmaceutical-specific biokinetic models for infants have been used [10Lei]. When such models are not available, biokinetic models for adults and iv injection have been used [00Sta]. As an alternative, radionuclide specific annual limit on intake (ALI)-values for adults corrected for body mass assuming a mass of the infant to 4 kg [94Rub, 89Mou] or 3 kg [90Ros] have been used.

4.9.3 Recommendations for interruptions

Since many radiopharmaceuticals are secreted in breast milk, it is safest to assume that, unless there are data to the contrary, some radioactivity of the radiopharmaceutical administered to a lactating female in whatever compound will be found in the breast milk. Consideration should be given to

postponing the procedure. If the procedure is performed, the child should not be breast fed until the radiopharmaceutical is no longer secreted in an amount estimated to give an effective dose >1 mSv to the child [87ICR, 09ICR]. Special concern should be given to iodine $^{123,125,131}\text{I}$ - and ^{75}Se - and ^{67}Ga -labelled radiopharmaceuticals as well as to ^{22}Na and ^{201}Tl . It is therefore recommended that the following actions should be taken for various radiopharmaceuticals.

Table 4.9.1. Recommendations for interruptions of breastfeeding after administration of radiopharmaceuticals [09ICR].

Radiopharmaceutical	Interruption	Radiopharmaceutical	Interruption
^{14}C-labelled			
Triolein	No	^{123}I -BMIPP	>3 weeks $\ddagger\ddagger$
Glycocholic acid	No	^{123}I -HSA	>3 weeks $\ddagger\ddagger$
Urea	No	^{123}I -iodo hippurate	12 h
$^{99\text{m}}\text{Tc}$-labelled			
DISDA	No *†	^{123}I -MIBG	>3 weeks $\ddagger\ddagger$
DMSA	No *†	^{123}I -NaI	>3 weeks $\ddagger\ddagger$
DTPA	No *†	^{125}I -HSA	>3 weeks \ddagger
ECD	No *†	^{125}I -iodo hippurate	12 h
Phosphonates (MDP)	No *†	^{131}I -iodo hippurate	12 h
Gluconate	No *†	^{131}I -MIBG	>3 weeks \ddagger
Glucoheptonate	No *†	^{131}I -NaI	>3 weeks \ddagger
HM-PAO	No *†	Others	
Sulphur colloids	No *†	^{11}C -labelled	No ¶
MAA	12 h	^{13}N -labelled	No ¶
MAG3	No *†	^{15}O -labelled	No ¶
MIBI	No *†	^{18}F -FDG	No
Microspheres (HAM)	12 h	^{22}Na	>3 weeks \ddagger
Pertechnetate	12 h	^{51}Cr -EDTA	No
PYP	No *†	^{67}Ga -citrate	>3 weeks \ddagger
RBC (<i>in vivo</i>)	12 h	^{75}Se -labelled	>3 weeks \ddagger
RBC (<i>in vitro</i>)	No *†	$^{81\text{m}}\text{Kr}$ -gas	No
Technegas	No *†	^{111}In -octreotide	No
Tetrofosmin	No *†	^{111}In -WBC	No
WBC	12 h	^{133}Xe	No
		^{201}Tl -chloride	48 h

* ‘No’, interruption not essential.

† ‘No’ for most of the $^{99\text{m}}\text{Tc}$ -labelled compounds, under the circumstance that no free pertechnetate exists in the radiopharmaceutical. An interruption of 4 h during which one meal is discarded can be advised to be on the safe side.

‡ 3 weeks (≈ 500 h) at least. However, difficult to maintain the milk supply, which often means cessation.

§ ^{123}I , all substances labelled with ^{123}I (except iodo-hippurate): >3 weeks due to the risk of contamination of other iodine isotopes.

¶ ^{11}C , ^{13}N , and ^{15}O -labelled substances, interruption not essential due to short physical half-life.

4.9.4 References for 4.9

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4.10 Current Trends and Future Developments

Dosimetry in Nuclear Medicine Diagnosis and Therapy

D. NOBKE, S. MATTSSON, L. JOHANSSON

4.10.1 Diagnostic imaging

As other imaging technologies, nuclear medicine/functional and molecular imaging develops fast. It has a uniquely high sensitivity for tracer concentrations compared to all other *in vivo* radiation based imaging techniques. PET can image and quantify tracer concentrations down to 10^{-15} mole/l, SPECT and planar gamma cameras down to 10^{-13} mole/l. This has to be compared with CT and MRI for which the comparative figure is 10^{-3} - 10^{-5} respectively (related to I and Gd-containing contrast agents, respectively). Currently, the fastest developments are within applications in oncology (both PET and non-PET techniques). Another field of development is in neurology and neuropsychiatry (with new tracers for amyloid plaques and for neurotransmission). Cardiology is also an area of expansion as surgery (imaging of sentinel lymph nodes) and urology (prostate cancer). In general, the developments in nuclear medicine are based on the developments of new radiopharmaceuticals that can visualise complicated intracellular molecular functions with very high sensitivity. This opens the possibility for individualised treatment planning and an early evaluation of therapy during treatment. Also the equipment develops. The introduction of hybrid imaging, e.g. SPECT/CT and PET/CT, – a combination of structural and functional or molecular imaging – has been an important step in nuclear medicine, contributing to better clinical handling of the patients. Patients with oncological diseases constitute the majority at a PET/CT department today, and around one third of the patients are treated differently when PET/CT images are added to the routine workup (using CT only) before treatment or follow up. For dosimetry, there is a need to make dose estimates for a number of new PET-substances. From the radiation dosimetry point of view, it should be kept in mind that PET/CT is a combination of two high dose investigations. Therefore it is important to have good dosimetry for a number of new PET-substances as well as for the CT-part of the investigations. In situations when enough biokinetic data are not available for a specific substance, generic models could be developed and if this is not possible realistic worst case examples could be given for substances used in nuclear medicine diagnostics [08ICR]. In the future, an automatic radiation dose report should be included along with the images when acquiring patient studies in the same way as is now more and more done for X-ray investigations. Such reports will be of interest for the individual patient and for site-to-site comparisons.

4.10.2 Therapy

In therapeutic applications of nuclear medicine/targeted therapy, there is a need for patient individual dosimetry with a high accuracy (see Sect. 4.7). At present, the established method for dosimetry for diagnostic as well as therapeutic purposes is based on measurement of the biokinetics by serial γ -camera images, followed by calculations of the residence times, resulting in the absorbed doses to tumours and critical organs per unit of administered activity. However, the quantification of the activity in different organs from planar data is hampered by inaccurate attenuation and scatter correction as well as influences of background and organ overlay. In contrast, dosimetry based on quantitative three-dimensional data can be more accurate and allows an individualized approach, provided that effects that degrade the quantitative content of the images have been corrected for. Matched anatomical imaging, such as combined SPECT/CT and PET/CT, has also made it possible to obtain tissue density information in conjunction with the radionuclide distribution. Coupled with iterative reconstruction algorithms, these advances have made it possible to perform highly patient-specific dosimetry (see, e.g., [09Sjö]). Advances in imaging will also increase the possibilities to evaluate the spatial distribution of

radionuclides within tumours and normal organs at various times after administration. The goal for radionuclide therapy should be the same as for external beam radiotherapy. Up to now less information has been reported on normal tissue tolerance and antitumor efficacy of radionuclide therapy than is generally available for external beam radiation. The correlation between reported dose estimates and biological effects has improved during the past two decades, partly as the result of increased accuracy and standardization of dosimetry techniques and to adjustment for biologic effects. With radionuclide therapy, the tolerance of normal tissues often appears greater but more variable. Much of the increased observed variability may be attributed to differences in dosimetry methodology and to heterogeneous distributions of the radionuclides, which complicates the estimates of the effect. In addition to that, factors such as overall treatment time and dose rate, radiobiological factors, genetic differences might be of importance.

Today there is also research going on to use radiobiological modelling to convert the spatial distribution of absorbed dose to a biologically effective dose (BED) [07Pri] for tumour tissue and for normal tissues.

In recent years, there has also been an increasing interest in combining biologically specific targeting agents (i.e., antibodies, peptides, etc.) with short-range particulate radiation emitters (alpha particles, beta particles, Auger electron emitters). This therapeutic combination offers the potential of delivering lethal doses of radiation to individual tumor cells while minimizing the volume of normal tissue irradiated. In these therapeutic applications, the dose needs to be determined on a scale that is comparable with the range of the particle emission. This scale is on the order of millimeters for beta particles, micrometers for alpha particles, and nanometers for Auger electrons. Heterogeneity of dose deposition at the cellular level will also remain of concern for the future. Both so called small-scale dosimetry and microdosimetry have up to now had limited applications in clinical practice. Accurate and complete small-scale dosimetry and micro dosimetry require knowledge of the source distribution as a function of time on the cellular/subcellular scale. In the clinic, activity determination is mainly based on quantitative scintigraphic imaging, i.e., with a spatial resolution on the order of mm to cm, and dose estimate done on organ levels. Calculation of absorbed dose on this scale has been sufficient for photon emitters used in imaging applications. At this scale, particle emissions are typically considered as nonpenetrating radiations, even when voxelbased activity determination is considered. In microdosimetry and small scale dosimetry, assessment of the geometric target is even more difficult as the target can range from single cells in suspension (i.e., ascites, blood-borne diseases) to small metastatic clusters to potentially macroscopic tumor masses. Clearly, the scale of target definition desired is much lower than available from current anatomical imaging methods (CT or magnetic resonance). It is a challenge to develop small scale dosimetry and microdosimetry for particle emitters for use in conjunction with cellular studies *in vitro* as well as *in vivo* studies in animals and later in man.

4.10.3 References for 4.10

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4.11 Summary and Conclusions

Dosimetry in Nuclear Medicine Diagnosis and Therapy

D. NOBKE, S. MATTSSON, L. JOHANSSON

Chapter 4 describes the current status of radiation dosimetry in nuclear medicine. Over the years, there has been an impressive development of both quantitative measurements and quantitative imaging as well as of the methods for absorbed dose calculations. However, the biokinetic behaviour of the tracer is still the most uncertain part of the assessment.

In the field of diagnostic nuclear imaging, as in diagnostic radiology, the reason for good absorbed dose estimates is the need to quantify the long-term risk of stochastic effects in the investigated individual, mainly manifested as an increased frequency of cancer later in life. Very seldom, the dose is so high that there is any risk for acute tissue effects.

In therapy with radiopharmaceuticals, the goal is to use dosimetry to foresee the response of a treatment both on the tumour tissue and on the normal tissue in the same way and with the same accuracy as in external radiation therapy. In external radiation therapy, it has been shown that basic absorbed dose data are good predictors of treatment effects. Further work will show if the implementation of radiobiological models will improve the prediction possibilities.

5.1 Introduction

Medical Radiological Protection

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Chapter 5 begins with a prefatory review of the background and development of radiation use in medicine and of radiological protection. After a summary of the current concepts and recommendations in protection against ionising radiation, it addresses the application of these protection principles to patients and to staff and members of the public in terms of clinical management, diagnostic and therapeutic radiology, and diagnostic and therapeutic nuclear medicine. Protection against non-ionising radiation as used for diagnostic purposes is also discussed, focusing on magnetic resonance imaging and diagnostic ultrasound. Finally, some of the safety aspects are reviewed.

5.1.1 Historical flashback

With Röntgen's discovery of x rays towards the end of 1895, the useful implications for diagnostic medicine were realised instantly. With a rapidity that is absolutely amazing by today's standards, X-ray diagnosis was introduced at hospitals and clinics, and more demanding applications were soon to follow – for instance, X-ray equipment was used at military field hospitals within two years [98Chu].

Given this rapid pace of introduction, it is not surprising that the possibility of deleterious side reactions was also acknowledged within months of Röntgen's discovery, with Grubbé's discovery of X-ray dermatitis (cf. 33Gru) and Drury's description of radiation damage to hands and fingers [96Dru].

Becquerel's identification of radioactivity in 1896 and Curie's isolation of radium in 1898 broadened the spectrum of medical uses of ionising radiation: Radioactive substances as well as x rays were soon used for radiation therapy, too.

However, during the first two decades of use of radiation, numerous medical workers suffered serious or even fatal consequences of radiation exposure, and the need for organised protection against this occupational hazard became obvious. In 1925, the first ever International Congress of Radiology (ICR) was held in London. At that time, the most pressing issue was to quantify measurements of radiation, and what is now known as ICRU, the International Commission on Radiation Units and Measurements, was formed. Three years later, the second ICR was held in Stockholm, where what is now known as ICRP, the International Commission on Radiological Protection, was formed, and published its first advice [28IXR].

The 1928 Recommendations focused entirely on occupational exposure in medicine, and the risks that were recognised were 'injuries to superficial tissues' and 'derangements of internal organs and changes in the blood' (i.e., what we would now call tissue reactions, or deterministic harm). The recommendations were written purely in terms of time, distance, and shielding, i.e., not in terms of any kind of dose limit. However, it has been estimated that the recommended limit on the number of hours worked with radiation may have corresponded to a dose limit in the order of 1000 mSv in a year [98Lin].

Six years later, the first radiological protection recommendations including a numerical restriction on exposure were provided [34IXR]. Those recommendations assumed a 'tolerance dose' (for staff working with x rays) of about 0.2 r per day. This would correspond to an annual effective dose of about 500 mSv, i.e., about 25 times the present annual limit on average occupational dose and about 10 times the present limit on occupational dose in any one year.

In the 1950s, more comprehensive recommendations on radiological protection emerged, with more restrictions. At first, the purpose of the added restrictions was to consider deterministic damage to more sensitive people. However, soon the focus turned to the probability of stochastic effects such as cancer and genetic damage. This meant that there was no obvious threshold dose below which radiation would be 'safe'. At low doses, a linear, no-threshold dose-response relationship could not be excluded, and indeed was widely perceived as a good approximation of the actual situation.

Therefore, it was no longer satisfactory to focus exclusively on occupational exposures, with their sometimes relatively high individual doses. Doses to individual members of the general public, and to individual patients undergoing medical examinations with radiation, were usually smaller than individual occupational doses, but the much larger populations concerned meant that the collective doses to members of the public and to patients were also much higher, and that therefore significant efforts were needed to provide adequate protection to these groups. The 1977 ICRP Recommendations [77ICR] were a milestone in this respect and are still very much reflected in the current ICRP Recommendations [07ICR]. Over the last 40 years, numerous ICRP reports have addressed protection of the patient; the most recent general overview of protection in medicine [08ICR] comprises both a summary of the general principles of radiological protection, adapted to the needs of the medical practitioner, and a brief review of recent topical advice concerning various specific problems in medical radiation usage.

5.1.2 Why radiological protection is essential

Indisputably, the introduction of ionising radiation as a tool in diagnostic and therapeutic medicine has led to an enormous improvement of human health. The benefit to the individual patient of taking a radiological examination will virtually always far exceed any risk associated with the investigation. Radiation therapy entails a higher risk, but is also life-saving. So why is radiological protection in medicine so important?

The answer is primarily that the focus has shifted from individual risks to collective risks to society. Furthermore, the main concern is now patients rather than medical staff.

If we start by looking back at occupational exposures in medicine, where the entire concept of radiological protection emerged, the situation has improved very much compared with the early years of diagnostic radiology. In the early years of the 20th century, many hundreds of medical practitioners were victims of serious and often lethal deterministic harm. Today, average annual effective doses to staff are very much lower, in the order of 0.5 mSv world-wide [08UNS]. Thus deterministic harm is prevented and stochastic harm is kept to a minimum (and certainly below the pertinent dose limits which can be seen as the border line, in terms of stochastic risks, between what is always unacceptable and what can in some circumstances be regarded as tolerable).

However, the number of staff concerned has increased considerably. The most recent world-wide data, from 2000-2002, is that radiation doses are monitored for about 7½ million medical staff, up from 2½ million in 1990-1994 [08UNS]. The annual effective doses to some limited groups of medical staff, although below dose limits and much lower than doses 50 years ago, can still be in the order of 5-10 mSv [08UNS], i.e., relatively high in comparison with other professions occupationally exposed to radiation.

Turning to what is now regarded as the primary concern, doses to patients, there is a tremendous range of individual dose, from essentially trivial doses due to dental examinations (effective dose in the order of 0.01 mSv or less; 08UNS), to highly significant doses outside the target area to cancer patients (effective dose in the order of a few hundred mSv; 85ICR). The number of patients concerned is enormous. Medical radiation comprises the largest man-made contribution to dose, and in some industrialised countries is now approximately equal to that of natural background radiation [08Met]. However, the collective dose is very unevenly distributed between countries, so while an over-utilisation in the shape of some unwarranted examinations can be suspected in some industrialised countries, much of the population in developing countries still needs improved access to medical radiation procedures.

Nevertheless, the huge number of patients world-wide means that the collective dose to patients is also huge (e.g., towards the end of the 20th century, the annual collective effective dose due to diagnostic X-ray examinations was in the order of 2.3 million manSv [00UNS] and has certainly continued to increase since then).

Computed tomography (CT) is a particular case in point. This procedure is extremely useful in medical diagnosis and therefore its use has increased tremendously over the last 25 years. However, CT is also a high-dose procedure, with individual effective doses to patients easily reaching and exceeding 10 mSv. In USA, there are currently 67 million CT procedures annually [08Met], causing an annual

collective effective dose of 440,000 manSv, i.e., an annual average per caput effective dose to the US population of 1.5 mSv.

Fluoroscopically guided interventional procedures constitute another important example. These affect fewer patients, but the dose to an individual patient can easily be much higher (i.e., far above any dose limit that would have applied to occupational exposures, and high enough to lead to deterministic harm [00ICR]).

Since the induction of cancer and genetic disease at low doses is likely to follow approximately a no-threshold linear dose-response relationship, this also means that there is a significant amount of iatrogenic stochastic harm. The exact amount is difficult to estimate, since patients differ in several respects from the general population, but the amount of detriment is large. It can be reduced, primarily by judicious use of radiation procedures such that unwarranted examinations are avoided, and by ensuring that radiological protection of the patient is optimised.

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5.2 Conceptual Radiological Protection and International Recommendations

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Within months after Röntgen's discovery of x rays, it was known that ionising radiation could cause serious skin burns; anecdotal evidence that x rays could also cause cancer appeared within a couple of years. The enormous potential benefits of radiation procedures in medicine were obvious; given the serious possible side effects, and the numerous early instances of serious harm to medical staff, it is not surprising that early on, the medical profession also preoccupied itself with devising concepts of radiological protection. As we have seen, the initial concern was for occupational exposures in medicine only, and the purpose of protection was seen only as preventing deterministic detriment to medical staff. Today, protection against occupational exposure in medicine is often (but sometimes overly optimistically) regarded as fully satisfactory, and the main focus is on exposures of patients and keeping stochastic detriment at bay.

5.2.1 The basic principles of radiological protection

In order to achieve the aim of preventing deterministic harm and minimising stochastic harm without unnecessarily impeding the beneficial uses of radiation, ICRP uses three basic principles [07ICR].

Justification

Any decision that alters the radiation exposure situation should do more good than harm.

Optimisation of protection

The likelihood of incurring exposure, the number of people exposed, and the magnitude of their individual doses should all be kept as low as reasonably achievable, taking into account economic and societal factors.

This means that the level of protection should be the best under the prevailing circumstances, maximising the margin of benefit over harm. In order to avoid severely inequitable outcomes of this optimisation procedure, and in order to take account of concurrent exposures from multiple sources, there should (in occupational and public exposure situations) be restrictions on the doses or risks to individuals from a particular source. Such restrictions are known as dose, or risk, constraints (in planned exposure situations) and dose, or risk, reference levels (in emergency and pre-existing exposure situations).

Application of dose limits

The total dose to any individual from regulated sources in planned exposure situations other than medical exposure of patients should not exceed the appropriate limits specified by the Commission.

5.2.2 Application of radiological protection to medical exposures

There are three general kinds of radiation exposures: occupational exposures, exposures of members of the public, and medical exposures. The vast majority of medical exposures refer to patients exposed for diagnostic, interventional, or therapeutic purposes. However, they also include exposures (other than occupational) incurred knowingly and willingly by comforters and carers such as family and close friends

helping in the support and comfort of patients undergoing diagnosis or treatment, and exposures incurred by volunteers in biomedical research projects.

In the contexts of occupational exposures of medical staff or incidental exposures of members of the public from medical radiation procedures, the three basic principles of radiological protection apply in full, and in the same manner as in any other radiation context. However, dose limits, and dose and risk constraints, are irrelevant for medical exposures of patients, since they may reduce the effectiveness of the patient's diagnosis or treatment, thereby doing more harm than good. Dose limits are also irrelevant for comforters and carers as described above and for volunteers in biomedical research; however, for these groups dose and risk constraints may be relevant [08ICR].

5.2.2.1 Justification of medical exposures

The emphasis is therefore on the justification of the medical procedures and on the optimisation of protection [08ICR]. Medical exposure of patients calls for a different and more detailed approach to justification than most other planned activities with radiation. The medical use of radiation should be justified, as any other planned exposure situation, but usually the medical profession has the primary responsibility for the process of justification (in other contexts, government or regulatory authorities usually have the primary responsibility).

Justification of medical radiation procedures applies at three levels: First (and this can nowadays be taken for granted), the use of radiation in medicine is accepted as doing more good than harm. Second (and this is usually judged by professional bodies in collaboration with pertinent regulatory authorities), a specified procedure with a specified objective is defined and justified, so as to judge whether the radiological procedure will usually improve the diagnosis or treatment or will provide necessary information about the exposed individuals. Third (and this is very important in the day-to-day work of the medical practitioner performing or requesting radiation procedures), the application of the procedure to an individual patient should be justified (i.e., the particular application should be judged to do more good than harm to the individual patient).

As pointed out by UNSCEAR [00UNS, 08UNS], several studies have highlighted the problem of unnecessary exposures. An analysis in the United Kingdom [90NRP] suggested that at least 20% of examinations were clinically unhelpful. Practical guidelines for the appropriate use of diagnostic radiology have been produced by international bodies, such as the World Health Organization [83WHO, 87WHO, 90WHO] and the European Commission [07CEC], as well as by national professional bodies such as the Royal College of Radiologists in the United Kingdom [07RCR, an updated and amended version of a document first published in 1989].

Thus, all individual medical exposures should be justified in advance, taking into account the specific objectives of the exposure and the characteristics of the individual involved. For high-dose examinations, individual justification is particularly important and should take account of all available information.

The process of justification should include checking that the required information is not already available and that the proposed examination is the most suitable method of providing the clinical information required. Avoiding indiscriminate referral is extremely important; if referral criteria and patient categories are defined in advance, usually this is neither cumbersome nor time-consuming.

5.2.2.2 Optimisation of radiological protection for medical exposures

In medical radiation exposures of patients, the benefits and detriments are received by the same individual, the patient, and the dose to the patient is determined principally by the medical needs. Therefore, dose constraints (the source-related restrictions on individual dose that are used to guide the optimisation of protection for occupational and public exposures) are inappropriate for optimisation of protection of the patient.

Nevertheless, some management of patient exposure is needed. *Diagnostic reference levels* constitute a useful tool as a mechanism to manage patient dose to be commensurate with the medical purpose in the

optimisation of protection for exposures from diagnostic and interventional medical procedures [96ICR, 01ICR].

Diagnostic reference levels apply to radiation exposure of patients resulting from procedures performed for medical imaging purposes. They do not apply to radiation therapy. Diagnostic reference levels have no direct linkage to the numerical values of dose limits or dose constraints. In practice, the values are selected on the basis of a percentile point on the observed distribution of doses to patients or to a reference patient.

Diagnostic reference levels are used to indicate whether the levels of patient dose from, or administered activity (amount of radioactive material) for, a specified imaging procedure are unusually high or low for that procedure. If so, a local review should be initiated to determine whether protection has been adequately optimised (i.e., the discrepancy reflects particular conditions for individual patients or groups of patients) or whether corrective action is required. The most common deviation consists of unusually high doses and the corrective action would be to reduce doses, taking care not to lose important diagnostic information. However, if the observed doses or administered activities are consistently far below the diagnostic reference level, there may be reason to conduct a local review of the quality of the images obtained.

ICRP has compiled a general overview of diagnostic reference levels [01ICR], but national and local professional bodies should be consulted since the selected values of diagnostic reference levels are often specific to a country or region. Extensive information is available on the management of patient dose in fluoroscopically guided interventional procedures [00ICR1], computed tomography [00ICR2], and digital radiology [04ICR].

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5.3 Applied Radiological Protection

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5.3.1 Management of radiological protection system in the clinic

Radiological equipment and radiopharmaceuticals are used in medical practice for a variety of diagnostic and therapeutic procedures. In order to carry out these procedures, different health professionals must be involved, e.g., medical practitioners referring patients to the examination or treatment, technical staff maintaining the equipment, staff performing the irradiation of the patient, physicians prescribing the treatment or evaluating the diagnostic examinations and medical physicists being responsible for dosimetry and dose planning. All these professionals must be involved and have to have the appropriate knowledge of radiological protection. It is important to distribute their duties and responsibilities so that the work is organised efficiently.

Technological development of medical equipment is fast. New technical solutions and accessories as well as new diagnostic and therapeutic procedures are frequently introduced in the clinic. This puts great demands on the staff handling the equipment and performing the procedures. Thus, a system including safety assessments before the introduction of new methods or equipment is necessary.

Diagnostic examinations and treatments are performed in different kinds of organisations, from small clinics perhaps possessing just a few equipment items and performing rather standardised procedures, to large university hospitals involved in the development of new techniques and procedures. Of course, radiological protection work differs very much between these different organisations.

Managing radiological protection [02IAE] includes ensuring that the basic principles (justification, optimisation, and the use of dose limits and dose constraints) are handled appropriately in practice. Management involves all levels in the organisation, including managers of clinics and departments, medical doctors, nurses, technicians, engineers, and medical physicists. All of this vast number of professionals has to be aware of their own role and responsibilities. Managers must allocate sufficient resources.

For instance, the disposal of discarded radioactive sources by sending them to a waste repository is often very costly. This must be planned for in advance, and a budget has to be allocated. Education and training also cost money and time; managers must recognise that these activities are important.

5.3.1.1 A Quality Assurance System supports radiological protection

Radiological protection should be considered as part of other quality assurance work, not as an isolated issue handled by a few experts. A proper Quality Assurance System, including all protection and safety issues, is an essential part of up-to-date management. Documenting diagnostic and therapeutic procedures used in the clinic is of course a core activity. Education and training of personnel is also a prerequisite for radiological protection and its organisation, recording and follow-up should be described as part of the Quality Assurance System. The quality control of equipment, taking into account protection and safety issues, is also best included in the Quality Assurance System of the clinic.

A Quality Assurance System requires commitment from managers and staff and requires teamwork involving different professions. The development and improvement of quality assurance systems, and thus of the quality achieved in the clinic, requires an organisation with a learning attitude, also with respect to mistakes – in short, a good safety culture.

High competence of the personnel is the most important aspect in order to ensure good protection and safety for the patients. The performance during examinations or treatments will certainly affect both the outcome of the procedure and the safety of the patient. All personnel using any of the equipment in the clinic need to know, to a certain extent, how the technique works, how the equipment should be handled,

and how to act in case anything goes wrong. The procedures applied and any changes must be explained and training must be provided to all personnel, and there needs to be a functional system to ensure this in the clinic. Documented procedures will be a substantial asset in this work.

5.3.1.2 Stakeholder involvement and communication skills are necessary

Other persons than those working with radiation in the clinic could have a legitimate interest in, and have opinions on, how examinations and treatments are performed. Such stakeholders include, e.g., patients, relatives, media, and politicians. The involvement of such stakeholders is subject to much discussion and their views are clearly needed, since neither protection nor safety is based on science alone – various value judgments are also involved.

Professionals in the radiological community must be able to communicate with stakeholders. As a complication, vast resources of information (correct or otherwise) now readily available to laymen could lead to misunderstandings about the risks and benefits of radiological procedures. Professionals in the medical sector must be able to, and are obliged to, communicate appropriately the risks as well as benefits of the procedures performed to others than their own community.

5.3.1.3 ‘Third parties’: Attention must be paid to others than patients and personnel

Obviously, patients and staff need protection and safety, but other persons also need to be protected and safe [[07ICR](#)].

Comforters and carers include relatives and other persons who voluntarily help and support the patient. Radiation doses to comforters span a very wide range. In diagnostic radiology, doses are easily kept to a minimum, but in nuclear medicine therapy, comforters and carers need proper instruction on how to keep their doses at an acceptable level (see [Sect. 5.3.6.2.5](#)).

Dose limits should not be applied to comforters and carers. For them, a radiation dose in excess of the dose limit for members of the public (1 mSv in a year) may well be justified and tolerable. Of course, optimisation of protection is still necessary. Unnecessary irradiation should always be avoided, for example by limiting the time spent with the patient, by using shielding between the patient and comforters, and by adjusting the distance. Special arrangements could be necessary when caring for children because they need more attention.

Doses in different situations must be assessed. General dose constraints for comforters have been suggested (see, e.g., [[97ICR](#)]) but other constraints for specific situations could be applied.

Exposure to volunteers in biomedical research also needs special attention [[98CEC](#)]. Clinical research projects are very common, e.g., when introducing new radiopharmaceuticals. Such clinical tests could involve either healthy volunteers or volunteer patients with a disease. Radiological equipment could also be used for other types of research projects involving volunteers (healthy and/or with a disease).

The volunteers involved have to be informed about the risk and agree explicitly to the exposure. It must be quite clear that this applies not just to healthy volunteers, but also to clinical research projects where the patient is also treated for a specific disease: the additional exposure has to be explained and agreed upon, and protection must be optimised for each situation so as to avoid unnecessary exposure. As far as possible, diagnostic examinations should be adjusted to meet the specific research needs only, e.g., in CT examinations the number of slices should be minimised. In all research projects involving humans an ethics consideration is mandatory, as stated in the Helsinki Declaration [[08WMA](#)].

The general public includes persons living, residing, visiting, or working near facilities using radiation, and those who could be exposed from discharged patients or waste. Possible exposure situations must be known in order to achieve proper protection of members of the public. The protection of the public must be taken into account in the design of facilities with, e.g., proper structural shielding. In order to keep doses at an acceptable level, restricted access and/or occupancy in- or outside facilities could also be considered. Monitoring of areas with public access may be necessary to confirm that protection is optimised.

Sometimes special attention is necessary. For instance, contractors working in the clinic may need to wear dosimeters and/or to modify their normal work procedures. Preset action levels could be established to aid such decisions. Dose records are very useful when dealing with later cases involving similar work procedures.

Finally, some exposure situations could lead to relatively high doses to the embryo/fetus [00ICRI]. For all exposure situations, regardless of whether the woman is a patient, a staff member, or a comforter, obviously proper protection of the child to be born requires that the pregnancy is known. Thus, the women concerned must be informed about the risks and encouraged/requested to tell appropriate staff members if they are, or could be, pregnant.

For pregnant personnel, the embryo/fetus dose must be assessed, taking into account potential exposures (i.e., the risk of accidental exposures). Such assessments should be prepared on a general level and permit general decisions in the clinic. These may need to be supplemented by individual assessments. Monitoring with an appropriately selected kind of dosimeter may be necessary (see also Sect. 5.3.2.3). An electronic dosimeter may be useful since readings can be taken frequently, even after each work procedure. Certain types of work may turn out to be unsuitable for pregnant staff members.

A pregnant comforter should be informed about the risks and special efforts should be taken to reduce the dose to a minimum and avoid exceeding the dose limit.

5.3.2 Facilities, equipment, and radiation sources

Designing and building a facility, or rebuilding an existing one, as well as purchasing equipment and accessories, requires proper competence [01IAE1] in order to address important issues such as safety and the potential to optimise procedures. Routines must be established in the clinic to ensure that these aspects are considered.

Medical equipment should follow ISO standards, but these standards only establish basic requirements, and different types and kinds of equipment still have different features. This must be taken into account when buying equipment. For instance, the technical specifications of digital image receptors used in diagnostic radiology will influence image quality and radiation dose. This must be known, and the appropriate requirements must be specified, in the purchasing procedure. Very often, medical equipment is controlled by computer software, and it must be ascertained in the purchasing process that such systems are safe. For example, such issues are of great importance in dose planning systems for external radiation therapy. Thus, proper expertise and involvement of the personnel that will handle the equipment are necessary when buying equipment.

The need for training should also be taken into account when new equipment and accessories are purchased.

New equipment must be checked before it is used on patients [97CEC]. Managers must be aware that sometimes this takes a lot of time, e.g., at least several weeks for radiation therapy equipment, and that too tight time schedules will increase the risk of accidents. It is important to decide in advance what requirements and checks have to be performed before the equipment may be used on patients, and to specify a maintenance plan for the equipment.

5.3.2.1 Specific issues concerning sealed sources

Ideally a *safety assessment* should be carried out before purchasing a radiation source [01IAE2]. Components of this assessment are: identification of situations where persons could be exposed, an estimation of dose levels in those situations (whether normal or abnormal), and identifying the need for training of personnel.

A *storage place* needs to be defined and security issues have to be considered. The storage place should only comprise radioactive sources and the place should be located where people are not unnecessarily exposed. The storage place should be protected from different environmental elements such

as fire. Of course, proper radiation shielding is required. These issues should be solved before the source has arrived to the clinic.

Planning of *disposal* of the source should begin when it is being purchased. It is important to remember that the life-time of the source is sometimes much less than the half-life of the radioactive material, due to the risk of leakage from old sources. Often the manufacturers do not guarantee the quality of the source shielding for more than approximately 10 years. It could be very costly to take care of a disused source, and if the life-time of the source has expired, this will most likely increase the difficulties with transport and disposal.

A *record of the sources* and where they are stored has to be kept. The condition of the source must be checked frequently. Wipe tests must be carried out in order to check that the sealing is not leaking. If a source is lost, the personnel should have access to monitoring devices suitable for the use in the search. The awareness of *security issues* has increased in the 21st century. High activity sealed sources as well as sealed sources with lower activity and open sources have to be protected from malicious acts, such as theft or sabotage. Proper security, including a limited number of persons having access to the sources is important. Hospitals and other health care units are often pretty accessible and a risk assessment must be performed for different types of situations in order to have an appropriate security. Thus, in some cases the source has to be protected from people rather than the other way around and persons with a proper knowledge of security have to be involved.

An *efficient work environment* at the workplace is a fundamental prerequisite for proper radiological protection. This requires the creation of efficient work procedures that minimize the risk for accidents. One example of this need for optimisation is the transport of sources that are used between storage and use areas, e.g., the handling of radiopharmaceuticals. The best route of transportation of the substances within the premises should be chosen in order to decrease the risk of contamination.

When choosing the *location* of the examination or treatment rooms, one should also look at the activities taking place nearby, not only at the same floor (building level) but also above and beneath the rooms. Critical areas, such as areas where many persons are present for a long time, have to be identified - these areas can cause problems. Future requirements also need to be considered when planning: changes such as introducing other types of equipment in a room could demand more shielding. The issues concerning radiological protection, including shielding assessments, have to be addressed early in the planning procedure. Again it is important to include the proper professionals such as nurses, medical physicists, physicians, etc., early in the planning stage.

5.3.2.2 Planning, verifying, and documenting structural shielding

In order to reduce exposure levels outside rooms and facilities, structural shielding in walls, ceiling, doors, windows, and floor could be required. The construction material and its thickness will define the exposure reduction. To a great extent, such structural shielding is needed for equipment used in diagnostics and therapy in order to properly protect both personnel handling the equipment and other persons working nearby.

Shielding requirements depend on: the type of equipment, the type of radiation, how much the source is used, the dose rate from the source, the direction of the beam, the distance from the source, the occupancy in the areas outside, the limit values in these areas, and what material comprises the structural shielding.

Shielding materials used for photons, gamma radiation, and x rays have a high atomic number; e.g., lead and iron will attenuate photons effectively. But for high-energy gamma and X-ray equipment used in external therapy, the thickness of the lead or iron needed will be very substantial, often too thick to use as structural shielding. Instead concrete is often used. In [Table 5.3.1](#), approximate tenth value layers (TVL) for lead and concrete for X-ray and gamma energies are given [[04NCR](#), [05NCR](#)].

Table 5.3.1. Approximate Tenth Value Layers (TVL).

TVL	Energy					
	50 kVp	100 kVp	150 kV	4 MV	10 MV	Co-60
Lead [mm]	0.17	0.88	0.99	53	55	40
Concrete [cm]	1.5	5.3	7.4	29.2	39.6	20.6

It is more difficult to arrange shielding against neutrons. The particle loses energy through collisions. The cross-section differs for different materials and is critically dependent on the energy. Light elements such as hydrogen, boron, water, and paraffin can be used as attenuating material.

Dose constraints commensurate with a reasonable level of occupational annual effective dose should be used in the planning process. Dose constraints for areas where radiographers, radiologists, etc., work are often set to 5 mSv in a year. In other areas, accessible to persons not working in the clinic, constraints are often set to 0.3 mSv in a year. This value is chosen to make sure that the dose limit for the general public is not exceeded. Note that in a hospital, personnel not involved in the work with radiation should have the same level of protection as the general public. This applies to, for example, the physiotherapist or the staff in the cafeteria.

The occupancy factor, how long time people are residing in areas outside the room, has to be assessed. Some suggestions on factors are given in [Table 5.3.2 \[05NCR\]](#). The occupancy factor could be modified by introducing local rules, for example a restriction on occupancy above a room when the equipment is used.

Table 5.3.2. Suggested occupancy factors in different areas.

Area	Occupancy factor
Full occupancy; control room, administrative and clerical offices, treatment planning room	1
Adjacent treatment room	0.5
Corridors, staff rest room	0.2
Storage areas, outdoor areas	1/20
Pedestrian area, vehicle traffic	1/40

Another important factor when assessing shielding requirements is the use factor: for how long time the equipment is used. An assessment of how the equipment is used must be included in the estimations. The primary beam could be directed towards the floor, walls, etc., and the details on the time in different direction must be included in the assessment.

The dose rate in the primary field as well as the dose rate due to scattered and leakage radiation must be known. Knowledge of the energy of the radiation is also important; for example leakage radiation will have a greater mean energy than in the primary beam. The presence of a patient will alter the primary beam substantially. This will introduce one more factor influencing the need for structural shielding.

The room size will also influence the need for shielding: theoretically, the larger the room, the lesser the need for shielding in the walls. The reason for this is of course the strong decrease of radiation levels with distance from the source. But note that the positioning of the source in the room also is very important.

It is obvious from the above discussion that the assessment will be fraught with considerable uncertainty. One also has to keep in mind that building procedures introduce further uncertainties. During the building process, the shielding and other important design features have to be checked, as it will be more difficult to check when the installation is finished. After the equipment has been installed, basic shielding verifications must be carried out to ensure that basic occupational safety requirements are met.

Further on, when the equipment is used on patients, the shielding should again be verified. The radiation dose could be measured over a period of time with normal workload in the clinic to ensure proper protection.

Sometimes it is necessary to re-assess shielding requirements when new equipment is installed in an existing room. On such occasions it is very useful to have documentation available of the calculations and assumptions made for the room in its present condition. This emphasises the need for documentation of the assessments of the shielding for future use.

5.3.2.3 Monitoring and classification of work areas

Monitoring the radiation environment inside the diagnostic or treatment room is helpful in numerous situations. Optimisation of work procedures [99IAE, 02IAE] requires the knowledge of exposure levels, e.g., finding the best position for staff participating in the procedure. Radiation doses to personnel not wearing dosimeters can be assessed and decisions can be taken on whether or not personal dosimeters are required.

Thus, all clinics should have access to equipment for monitoring radiation in and outside the facilities [04IAE]. Survey meters such as ionisation chambers and Geiger-Müller counters are useful. The instrument specification, e.g., energy response and sensitivity, must be studied in order to understand the outcome. Instruments could be calibrated for different quantities, e.g., ambient dose equivalent, $H^*(10)$ - an SI unit [93ICR] relevant to radiological protection - or counts per second. The latter instrument would be useful to detect radiation or to verify that the radiation level is normal. *Measurements must be performed by personnel with a thorough knowledge of the subject in order to interpret the results correctly.*

In order to indicate safety needs in a room or work area, a system of classification of different areas is frequently used in the regulatory system. The system includes supervised and controlled areas (or an explicit decision that a classification is not needed). A *supervised area* is a defined area for which occupational exposure conditions are kept under review, although no specific protection measures or safety provisions are normally needed. A *controlled area* is a defined area in which specific protection measures and safety provisions are, or could be, required for controlling normal exposures or preventing the spread of contamination during normal working conditions, and preventing or limiting the extent of potential exposures. A controlled area is often within a supervised area, but need not be.

The activities performed, the type of equipment, and the sources used in the room will determine the classification. Rooms used for therapy or for interventional procedures with x rays, and laboratories for the preparation of radiopharmaceuticals, are commonly classified as controlled areas. Others areas such as ordinary X-ray rooms are often supervised areas. Typical differences between supervised and classified areas are accessibility to the area, information required for persons entering the room, and the frequency of training activities such as emergency training. Signs stating the classification and type of sources involved and other safety messages are often required. In both areas, some level of access restriction must be defined.

5.3.3 Occupational exposure

Knowledge, competence, and ability are indispensable prerequisites of ensuring a high level of protection for personnel. This includes, amongst other things, the safe and proper handling of the radiological equipment.

The exact level of competence required is not easily defined, since this includes basic professional education, specific knowledge of radiological protection, and knowledge of local procedures used in the clinic and equipment specific features. The basic education of health professionals covers different issues in different countries. Often, additional courses are required; sometimes, these are mandatory before beginning to work in a new workplace. On-the-job training is also necessary, because different clinics

possess different types of equipment and the applied methods could differ substantially. Continuous education is also needed for most health professionals.

Thus it is hard to set up a general education plan in the clinic. Some parts have to be adjusted for the different groups of personnel dealing with different procedures and equipment. In such discussions, medical physicists are often overlooked; although basic radiological protection should be part of the basic education, this group of personnel also needs continuous education, and some medical physicists elect to specialise in radiological protection. In some countries, health physicists take care of certain tasks in radiological protection, e.g., shielding assessments and occupational protection, while medical physicists focus on the protection of the patients. If there are no health physicists, often the medical physicists will take care of the whole field of radiological protection.

5.3.3.1 Occupational doses

It is hard to get a good overview of occupational doses in medicine, because assessment techniques, the number of workers monitored, and the principles of collection of data vary within and between countries. However, some information is available, e.g., through the European study on occupational radiation exposure (ESOREX) in the European Union. According to ESOREX, the average annual occupational whole body doses in the medical sector remained rather stable through the early years of the 21st century, with an average level of the order of about 0.5 mSv per year [10ESO].

When studying such data it is important to remember that a large group of personnel do not get any dose at all, and in many cases are not expected to get any dose because of the nature of their work. For instance, the annual dose to a radiographer carrying out conventional examinations and exiting the examination room when exposing patients should be close to background levels. If these near-zero values are excluded, the resulting average dose to measurably exposed medical personnel is about 1 mSv per year.

These average values of occupational dose conceal considerable variation within the group of medical personnel. National data from the UK [05Wat], Fig. 5.3.1, shows that the average dose in nuclear medicine is about four times that in diagnostic radiology or radiation therapy.

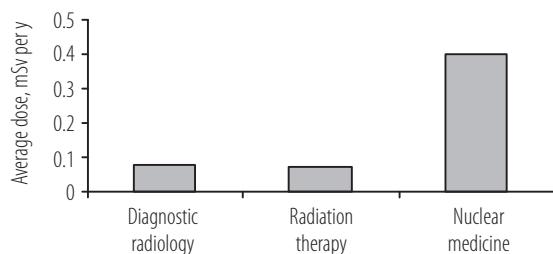


Fig. 5.3.1. Average annual occupational whole body dose (measurably exposed personnel only) for different medical sectors reported in the UK [05Wat].

Some activities in medicine, such as interventional radiology, can give rise to much higher doses, up to an order of magnitude higher than the average values mentioned above (see, e.g., 00ICR2). For some jobs and certain types of examination this may be true even if the work is performed under optimised protection conditions and protective equipment is used.

Furthermore, in some situations the dose to a specific body part is much higher than the average dose to the whole body [08Van]. For example, in some cases the hands are positioned very close to the radiation source and thus particular concern is required. Another example is the dose to the lens of the eye: for some procedures, the head is very close to the source and the dose to the lens must be assessed for some workers. When handling open sources, such as radiopharmaceuticals, the personnel could also be exposed through inhalation or ingestion [99IAE2]. The radiation dose depends on the type of radiopharmaceutical, the amount of activity, and the physical half-life as well as the biological half-life; the latter depends on the biokinetics of the radiopharmaceutical. Thus in medicine both whole-body dose and dose to specific body parts from external exposure as well as in some specific cases the radiation dose from internal exposure must be considered.

The occupational dose depends on a number of factors. The most important one is how the work procedures are planned and how well the procedures are performed; that is, optimisation of protection and training. Planning includes investigating the best use of personnel radiation protection equipment such as lead aprons, protective gloves, moveable shields etc. Performing the examinations and treatments efficiently will not only reduce the radiation dose, the risk of accidents will also be minimised. Unless activities are planned and education is provided, the dose limits could be exceeded in numerous activities, e.g., when using X-ray fluoroscopy equipment.

Another factor would be the specific workload at the particular clinic and for a particular worker. Thus, risk assessment and the planning of occupational dose management require deep knowledge of the work activities. The occupational dose in a medical clinic can not be estimated based on published values; it has to be assessed in every clinic.

5.3.3.2 Monitoring of occupational dose

Sometimes it is necessary to monitor radiation dose to workers individually. Effective dose and equivalent dose, the dose quantities that are used in regulatory systems, are not measurable quantities. For *external exposure*, a substitute is to determine the personal dose equivalent, $H_p(d)$ which is a so-called operational quantity [93ICR]. Three depths (d) are used, viz., $H_p(10)$ for estimating whole body dose, $H_p(0.07)$ for estimating skin dose, and $H_p(3)$ for estimating the dose to the lens of the eye (further reading Chapter 3 and [05Lan1]).

The personal dosimeters used must be calibrated in these quantities. The dosimeters could be provided by a dosimetry service. Before using such a service, the quality of their work should be checked. For most relevant radiation fields in the medical sector, the recorded personal dose equivalent will over-estimate effective dose and equivalent dose. Particular care is required in situations where neutrons or heavy charged particles are present.

Several types of dosimeters are available and all have specific characteristics and features [05Lan1]. This will result in uncertainties specific for each exposure situations [09CEC]. One commonly used type of personal dosimeter, the thermoluminescent dosimeter (TLD), is a so-called passive detector where the result is available only after an evaluation procedure. Sometimes it is desirable to be able to take immediate readings, e.g., in a training situation. If this is the case, it is preferable to use an active dosimeter. On such an instrument, the dose value will show immediately, or after a simple evaluation procedure that can be performed locally. For measurements of dose to skin and eyes, both types of dosimeter are available.

The position of the dosimeter on the body influences the measurement. Usually, the dosimeter is placed on the trunk for estimation of the whole-body dose. If a lead apron is used, this is usually simply positioned over the dosimeter. However, in some countries it is common practice to place the dosimeter outside the protective shielding. For practices with a heavy workload, it is sometimes recommended to use two dosimeters: one on the trunk and one above the apron, and then to use a weighted combination of the two readings for a better estimation of the effective dose [09Kui]. This will take into account both those body parts that are covered and those that are not covered by the shielding.

The positioning of dosimeters assessing the dose to particular body parts, skin, or lens, is also crucial due to the huge variation of dose levels. For example, placing a dosimeter at a surgeon's wrist will underestimate seriously the dose to the skin of the hands because the fingers are more likely to be in the primary radiation field.

For estimations of *internal exposure* a different approach has to be taken. The ingested/inhaled radioactive substance could contribute to radiation dose for a long time. Therefore, the quantity that should be assessed is the committed effective dose [07ICR] and this value depends on the effective half-life of the substance in the body. If the radionuclide concerned is a photon-emitting isotope, a whole-body counter could be used [05Lan1]. This is a device containing one or several detectors, and after calibration the activity in the body could be estimated. If this is not feasible, samples of substances such as urine or blood from the person could be examined. However, such measurements are not easy to perform, and in order to estimate the whole-body dose, some kind of calibration factor taking account of the biokinetics must be applied. Workplace monitoring of contamination on surfaces or in air could also be used to assess

possible radiation doses to the workers. Assumptions regarding, e.g., the total amount of intake will be required, and coefficients specific for different substance must be used in order to calculate the organ doses. Such estimates are always fraught with considerable uncertainty.

The purpose of dose measurements and assessments is not just to verify that dose limits are not exceeded, but primarily to provide an important input to the organisation of work practices in optimisation of protection. Personnel should be divided into three categories, viz., those whose personal doses are to be monitored continuously, those whose doses are reviewed occasionally, and those for whom no dose review is necessary. There must be a programme for individual monitoring [[99IAE3](#), [04IAE1](#), [04IAE2](#)] in the clinic, describing the allocation of personnel to these three categories.

It is important of course to evaluate doses as soon as measurement results are available. It is useful to keep records of measurements in order to follow dose trends. Usually, national law describes cases where the retention of records of individual monitoring is mandatory. Often, there is a national co-ordinated registry of individual doses.

Investigation levels are helpful in the interpretation of dose measurements and management of personnel doses. When an investigation level is exceeded, the reasons should be examined, with a view to revealing changes of work procedures or lack of good work procedures. The investigation level must be chosen such that it indicates if a measured dose is higher than normal; thus the normal level for different groups has to be known. Also, comparing occupational doses with other clinics is a very useful element in optimisation of protection.

Special arrangements could be necessary when a staff member is pregnant [[00ICR1](#)]. The special dose limit, 1 mGy, to the child to be born has to be met. Area monitoring could be of great help when investigating if special arrangements are required so as not to exceed limits and to minimise the dose. Personal monitoring with frequent evaluation of the dose received, e.g., using active dosimeters, could be valuable in this situation.

5.3.4 Systems for optimisation and justification

The optimisation of radiological protection in medical examinations and treatments includes investigating and evaluating a large amount of information in terms of technical and operational parameters. A systematic approach is necessary, and a sufficient amount of time must be allocated to working with optimisation.

In radiation therapy, optimisation is an integrated part of the treatment process. The treatment result at an aggregated level, viz., survival rate of the patients and frequency of secondary effects, reflects the quality of the optimisation.

In diagnostic examinations, it is much harder to evaluate the level of optimisation. For example, it is not easy to assess image quality in the clinic, or whether this quality reflects an appropriate level of the trade-off between dose and diagnostic requirements. Also, it is not easy to evaluate whether a radiation dose to a patient is reasonable, because the dose varies significantly between different diagnostic procedures: different body parts are examined for different medical conditions and need different doses in order to get useful information. Different procedures will be of different complexity and need different numbers of images or fluoroscopy times. The radiation dose will differ between different patients for the same examination, due to the fact the patients are of different sizes. Furthermore, there may be medical reasons why the same investigation needs to be performed with a higher dose to obtain more detailed information for some patients. Actually, the same dose to all patients for the same procedure would indicate poorly optimised protection. But normally, the radiation dose for the same examination for the same size of patient should not differ very much.

5.3.4.1 Diagnostic reference levels

Diagnostic reference levels [07ICR] for specific procedures have been introduced so that radiation dose can be compared for a given type of examination in different hospitals. If doses at a given clinic are consistently different than those at peer departments, an analysis of the causes is warranted. Such a deviation does not necessarily indicate any problems with optimisation; for instance, higher average doses would be expected at a clinic catering primarily for obese patients. However, it is important that the cause of any difference is known and that the difference is maintained only if appropriate in view of the conditions. Doses that are unusually low also require investigation; they may reflect insufficient image quality.

The diagnostic reference level is defined for a region or nationally and is used as a standard against which to compare radiation doses determined locally in the clinic. The concept of diagnostic reference levels has been introduced in both diagnostic radiology and nuclear medicine [99CEC]. Different strategies have been applied in different countries and different quantities for describing the radiation dose and different levels have been chosen. For example, the dose-area product can be used for conventional X-ray examinations and the administered amount of activity for nuclear medicine procedures.

To a large extent, diagnostic reference levels were established from national surveys where the range of existing doses was studied and the third quartile of the distribution has often been chosen as the reference level [09Har]. The scope of diagnostic reference levels is thus to minimise the amount of unnecessary high dose and to minimise unwarranted differences between hospitals [07ICR]. This is illustrated in Fig. 5.3.2 with a hypothetic example, the examination rooms with the highest dose are identified and actions to lower the dose are accomplished in these examination rooms.

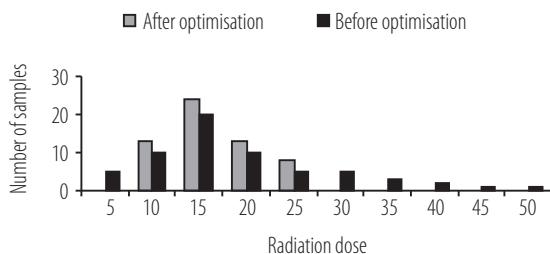


Fig. 5.3.2 The effect, schematically, on radiation dose after reference dose levels are applied in different hospitals.

Diagnostic reference levels can provide a starting point for optimisation, but in order to optimise protection, it is not sufficient simply to ensure that doses are below the pertinent DRLs. The image quality must also be taken into account. This is not a trivial task; research studies on image quality performed locally in a hospital are not easily applied generally to other hospitals.

5.3.4.2 Justification of diagnostic examinations

No radiological examination should be performed in the clinic unless it is justified, and this must be ensured for each patient. When referral of a patient is considered, it has to be evident that the outcome of the examination will contribute to the treatment of that individual patient. This implies that the referring physician and the radiological specialist should decide jointly on any radiological procedure on the basis of; the patients' case history, clinical examination of the patient, and the availability of radiology procedures or alternative techniques. To accomplish this, the referring physician and the radiologist should work together.

In order to facilitate the justification process and avoid waste of time, guidelines are used to handle frequently occurring situations and common patient types. In practice, the production of such guidelines is quite complicated. The efficacy of each procedure has to be evaluated and compared with other procedures. An individual clinic or hospital is not capable of evaluating each procedure in isolation. Therefore, international guidelines have been developed [07CEC], and each department has to create a

strategy to make the best use of these guidelines. Sometimes, such guidelines must be adjusted to local conditions or regulations. To some extent, the procedures chosen depend on the equipment available in the clinic or the availability of trained staff for a specific procedure. It is always very important that a clinic is aware of why a certain procedure is being chosen.

Applying procedures for optimisation and justification is especially important for children as children are more sensitive to radiation [07ICR]. Thus the benefit of some high dose examinations should be carefully weighed against the increased risk. The justification of examinations or treatment in pregnant women also requires special consideration due to the higher sensitivity to radiation of the child to be born [00ICR1].

The justification of screening activities poses particular challenges [07ICR]. Mass screening of population groups is not justified unless the expected advantages for the individuals examined as well as for the population as a whole are sufficient to compensate for the economic and social costs, including the radiation detriment. In principle, the diagnostic methods used have to be sufficiently sensitive to detect disease and specific enough not to indicate healthy individuals as having the disease. Above all, society has to be able to start a treatment of the individuals with disease.

5.3.5 Diagnostic and interventional radiology

5.3.5.1 Design of installations and structural shielding in diagnostic radiology

A simple and fairly easy way of protecting people around an X-ray source is to surround it by shielding material; it is easy to build in protection in the ceiling, floor, and walls. With sufficient shielding there will be no need for inconvenient restrictions on access to areas around the rooms. The shielding needs will differ for different kinds of equipment used in the room. Basically, the higher tube voltage and the more work load, the more shielding is required (although there will be additional considerations).

An important feature of radiological equipment with implications for shielding requirement is whether a primary beam could reach the shielding or not. In mammography, computed tomography, and fluoroscopic and interventional equipment, X-ray tubes and image detectors are fixed so that only scattered and leak radiation from the X-ray tube can reach the shielding. But if the workload is heavy, as for example in interventional radiology, the amount of scattered radiation is significant and structural shielding is still required.

Sometimes, shielding is required in windows and doors. This can be achieved by using special lead glass windows and lead shielding in the doors. The building technique is very important in order not to create any gaps in the shielding. Thus, for such work, experienced suppliers and contractors must be engaged.

General parameters affecting shielding requirements, such as workload as discussed in Sect. 5.3.2.2 above, must be estimated, but the shielding calculations are fairly easy to perform [04NCR]. All calculated shielding requirements are given with uncertainties that must be taken into account when evaluating the results of calculations. The building technique also influences the shielding; e.g., lead shielding is available in certain thicknesses only.

5.3.5.2 Optimisation of the protection of the patient in diagnostic radiology

A number of technical parameters will influence the radiation dose in diagnostic radiology directly, as discussed in earlier sections. A number of them are defined by equipment-specific features such as the image receptor, anti-scatter grid, and table-top attenuation. Furthermore, accessories such as computer programs used, image reconstruction, or image manipulation techniques will influence the conditions that must be taken into account in optimisation of the procedures. *This implies that the optimisation procedure starts when equipment is bought.*

In the clinic, a number of technical factors influencing the dose are often preset for different examinations, or parameters are automatically controlled during the examination. This includes factors such as tube voltage, tube filtration, source-to-skin distance, tube current, and dose rate. All these parameters have to be decided for each type of examination. This is the joint responsibility of the radiographer, the medical physicist, and the radiologist. *This implies that an important part of the optimisation procedure takes place prior to the clinical introduction of a procedure.*

Procedural parameters, such as number of images per examination or volume to be covered in computed tomography, are decided during the performance of the examination. Other parameters such as field size, possible compression of the body part examined, or the use of various shielding devices such as gonad shielding, will determine the final dose to the patient. This is often the radiographer's responsibility. *This implies that a significant part of the optimisation procedure includes education and training of the personnel.*

As mentioned above, both technique factors and procedural factors will influence the dose, but they will also influence image quality. The quality of the images also has to be addressed and specific situations using different techniques must be researched. Such an optimisation process is not easy to perform in the clinic; it requires a systematic approach comparing different examination techniques. Benchmarking with other clinics is a prerequisite for a successful optimisation process.

When using diagnostic reference levels in diagnostic radiology it is important that the radiation dose is assessed in an exact manner in order to be comparable with the designated level for the procedure. It must be kept in mind that as yet, diagnostic reference levels exist for a few examinations only. Ideally, the radiation dose should be assessed and evaluated for all kinds of examination.

An important part of the optimisation procedure, sometimes not mentioned explicitly, is the performance of the evaluator, in many cases the radiologist. The evaluation conditions, such as the quality of the computer screens, are important, but also, the quality of different types of computer databases where information on the patient as well as previous examinations are available influences the quality of the outcome.

It is worth mentioning that new technology does not guarantee an efficient image quality and low radiation doses. For example, a digital detector can be exposed to a very broad range of different radiation doses, all resulting in a sufficient image quality. The availability of computer manipulation can complicate the optimisation process further, so this is yet another factor to take into account [04ICR1].

5.3.5.3 Special considerations in diagnostic radiology

Some special cases posing particular challenges are detailed below.

5.3.5.3.1 Interventional radiology and fluoroscopically guided procedures

The scope of interventional and fluoroscopically guided procedures varies. Therefore, the use of the equipment will also differ. Thus, the need for using fluoroscopy and the number of exposures will differ between different patients, resulting in a varying patient dose as well. The large expected variation of dose combined with the rather low number of patients for each specific examination makes it difficult to compare radiation doses. However, it is important to perform dose surveys. Such surveys have shown that for such procedures, the radiation dose to individuals can be in the order of 20 – 50 mSv, and sometimes reach even higher levels [00ICR2].

Not just the whole-body dose is of concern for these procedures. The skin dose and lens dose must also be included in dose assessments. The skin dose and lens dose can exceed the threshold for deterministic injuries. Indeed, skin injuries have been reported, but due to the difficulties in identifying them, one can assume that the frequency is underestimated.

The patients who are at risk of developing skin injuries should be identified in immediate connection with the procedure performed [04COU]. Therefore, clinical action levels could be established, indicating when skin or lens dose should be assessed and documented and when the patient and the physician caring

for the patients should be informed. Then both patients and physicians can be observant and measures can be taken to relieve any harm caused by a radiation injury. Sometimes, it is necessary for important diagnostic reasons to continue to perform a procedure even though it is likely to result in a high radiation dose to the patient. However, injuries should not be incurred due to bad equipment, sloppy performance, or lack of training of the personnel, all of which are possible reasons underlying high doses.

A lack of proper education and training of the personnel involved in interventional radiology and fluoroscopically guided procedures has been identified and joint efforts have been taken to improve the situation [00CEC]. Thus special programmes must be established in the hospital, and this concerns introduction courses as well as continuous training programmes. A medical physicist, a radiographer, as well as a radiologist are all needed to set up and run such a programme. Last but not least, the manager of the specific department concerned must be aware that the programme is needed.

5.3.5.3.2 Computed tomography (CT)

Special attention should be given to the use of computed tomography. The number of investigations is increasing and the dose to the population from medical exposures today is dominated by CT investigations [09NCR]. It is possible to increase the number of examinations per unit time and operator due to technology development, more complex examinations can be performed with the new technology, and the development is rapid [00ICR3]. This calls for a particularly careful consideration of the justification of CT examinations.

New CT machines could either increase or decrease the radiation dose per examination [07ICR], depending entirely on how the examination parameters were set up. More complex procedures are being introduced, such as coronary angiography. An optimisation procedure taking image quality into account is needed. However, no standard procedure has been developed and fundamental image quality factors, such as noise, are known not to be useful on their own. Computed tomography examinations must be given particular attention, and optimisation work should be given priority for CT examinations.

Automatic exposure control is now commonly available and it is important to calibrate this device in order to optimise the dose. Two possible reasons underlying unnecessarily high doses are lack of current adjustment for different body sizes and increased irradiation volume in the body.

5.3.5.3.3 Procedures performed outside the X-ray department

Diagnostic radiology equipment is used in a variety of procedures outside the X-ray department and professionals other than radiographers and radiologists are using the equipment, for instance cardiologists, pulmonologists, gastroenterologists, orthopaedic surgeons, and dentists. These professionals are specialists in their own areas and usually have little or no knowledge about radiation or radiological protection. However, radiation protection issues are still very important and the equipment and procedures still have to be optimised. This requires advice from colleagues with a more profound knowledge about diagnostic radiology equipment and procedures. It is very important that this is handled in the clinic, setting up a team for optimisation and training in those departments.

5.3.5.3.4 Children

Fewer examinations are performed on children than on adults. This is one of the difficulties with paediatric radiology, because there will be less experience of performing these examinations – especially in hospitals not specialised in health care for children. Therefore, it is advisable if a dedicated group of personnel can be assigned to perform paediatric examinations.

The radiation dose will vary due to different body sizes and the dose assessment will be difficult. In some cases, the doses are so low that the measurement equipment will perform with great uncertainty. This could be a problem for example when measuring dose in the neonatal department.

The X-ray equipment must be checked in order to ensure that it will perform appropriately for paediatric examinations. In many cases, the exposure time will be much less than that for any examinations performed on adults. The use of the automatic exposure control will have to be adjusted to the smaller body. The necessity or otherwise of using a grid should be carefully considered, since a significant dose reduction can be accomplished if the grid is removed.

The procedures should also be adjusted for less co-operative children. Lung examinations, for instance, must be performed in a lying position until the child is big enough and sufficiently co-operative to stand up. Various immobilisation devices are also used in order to keep the child motionless so as to reduce the number of exposures and thus the dose.

5.3.5.3.5 Pregnant patients

Radiation doses to the child to be born are generally low in X-ray examination procedures. The risk of relatively high doses is greatest for interventional procedures in the lower abdomen. For procedures where the abdomen is not in the primary radiation field and only scattered radiation can expose the child to be born, such as lung examinations, the dose will be on such a low level that no restrictions on examination of the woman is needed. [Table 5.3.3](#) gives approximate radiation doses to the child to be born for some X-ray examinations [00ICR1]. Note that radiation dose varies very much between hospitals thus these doses will also vary.

Table 5.3.3. Approximate radiation dose to the child to be born in diagnostic radiology.

X-ray examination	Radiation dose [mGy]
Conventional x ray, chest	< 0.01
Conventional x ray, pelvis	1.1
CT, chest	0.06
CT, pelvis	25

However, unintentional exposure of pregnant patients should be avoided and for all X-ray examinations involving the lower abdomen, pregnancy should be excluded before proceeding with the examination. If the woman is pregnant, it should be carefully considered whether the examination is justified. Alternative diagnostic procedures not involving ionising radiation, such as ultrasound examinations, could be considered. If an X-ray examination is inevitable, protection should be carefully optimised. Measures to be taken could include reducing the number of radiographs, or reducing the irradiated volume in computed tomography. When relatively high doses are expected, such as in fluoroscopic examinations and interventional radiology, the optimisation should include an advance assessment of absorbed dose.

Sometimes pregnancy is discovered after an X-ray examination has been performed. In such circumstances, usually it would be sufficient to assess the radiation dose. It is very unlikely that an X-ray examination will cause a radiation dose to the embryo or fetus exceeding 100 mGy, and it would not be justified to terminate a pregnancy because of such a radiation dose. The clinic should be able to perform dose assessments and to communicate risks to the patients.

5.3.5.3.6 Carers and comforters

In diagnostic radiology, the protection of carers and comforters does not pose any great difficulty. If it is necessary for such a person to be present in the examination room, they could easily be positioned in a favourable place and protected with a shield or protective clothing.

5.3.5.4 Protection of personnel and the general public

The radiation doses received by personnel in diagnostic radiology vary greatly. Those personnel who never enter the examination room will not receive any absorbed dose at all if the room is properly designed. A large part of the staff performs that kind of work. On the other hand, unless proper working procedures are applied and protective equipment is used, the risk of exceeding present dose limits can not be excluded for personnel performing procedures such as fluoroscopy and staying very near the patient during exposure.

The exposure of a person is not homogeneous; knowledge of the radiation field is important to identify those personnel and procedures that need the most attention in order to optimise protection and to keep the doses to personnel as low as reasonably achievable. Procedures have to be developed, communicated, and trained.

Routines when entering the examination room should be established. Sometimes, the patient must be accompanied by radiology staff, staff from other departments, or comforters during the examination. One important issue is the use and availability of protective shielding, the effect of which when properly used is large. Clear instructions concerning how and when to use such equipment should be available. A variety of different types of protective shielding is available commercially. Aprons with at least 0.25 mm lead could be used in general radiology. In fluoroscopy, especially interventional procedures, the tube voltage may be greater resulting in a greater total dose. For such work, an apron with 0.35 mm lead is appropriate. Other types of protective equipment include, e.g., glasses and gloves.

It is important that it is agreed when to use protective equipment which is sometimes perceived as inconvenient and therefore easily ‘forgotten’.

Lead curtains hanging down from the examination table could also be used when there is an under-couch X-ray tube. This will reduce the dose to the legs of the operator to a minimum. Some shielding devices could be mounted in the ceiling.

All departments should examine what kind of devices could and should be used and decide when it is compulsory to use them.

5.3.5.4.1 Special considerations: interventional and fluoroscopic procedures

An assessment of the whole-body doses to personnel is necessary of course. In diagnostic radiology, particular attention should be given to ensure that radiation doses to the lens and the skin are not unnecessary high [00ICR2]. These investigations include procedures where the operator stands close to the patient and the procedures require hands to be very near or in the X-ray field, as well as when the eyes of the operator are close to the patient. Regular monitoring of radiation dose to the skin and lens is often required. Typical procedures are angiographic examinations and most interventional procedures.

It could be difficult to carry out measurements every time a person performs a procedure, but measurements should be performed in such a way that the yearly dose for each operator could be assessed with sufficient confidence. Thus, the radiation dose should be measured for a number of repetitions of any particular procedure and this information should be combined with knowledge of the total number of procedures performed in a year to permit an assessment of the operator’s annual dose. Special attention will be required of course when new procedures are implemented in the clinic.

The measurements could be performed using TLD dosimeters, specially protected in order to be carried under latex gloves. This is achieved by placing the dosimeter in a plastic ring or in a special plastic cover. Dosimeters not placed on the fingers will under-estimate the finger dose significantly.

5.3.6 Diagnostic and therapeutic nuclear medicine

5.3.6.1 Designing facilities and structural shielding: special considerations in nuclear medicine

The work in a nuclear medicine department includes a host of very different procedures including, e.g., laboratory work to produce the radiopharmaceutical required, administration of this radiopharmaceutical to the patient, restrooms for patients, examination of the patient, patient care, and storage of radiation sources as well as waste. Furthermore, the radionuclides used are transported to the clinic and between the rooms within the clinic.

Therefore, the design of the premises is very important and a *workflow analysis* should be used in order to optimise radiological protection. In this analysis, the areas where the highest radiation doses occur can be identified. The general layout of the nuclear medicine department should take into account a possible separation of the work areas and the patient areas. The transport of activity should be kept to a minimum. A *safety assessment* [05IAE] should be performed in order to determine different hazards. High-risk activities and areas would typically be the production of nuclides, the preparation room, and temporary storage of waste.

The laboratories where radiopharmaceuticals are prepared must be planned for the special needs in terms of ventilation, plumbing, and materials used in walls, floors, and workbenches. The sink where aqueous waste is released is best attached to the main sewer of the building. Materials could be contaminated and thus need to be easily cleaned. Worktop surfaces must be washable and chemical-resistant with all joints sealed. Fume hoods are used to reduce airborne substances and thus the filters in the ventilation system are contaminated; hence procedures for maintaining ventilation and to take care of these filters must be established. A portable contamination monitor must be available in the laboratory to render frequent monitoring possible, particularly of hands and sleeves.

Because of the potential for contamination of personnel working in the laboratory, arrangements to permit staff to wash are important. For example, a washbasin should be located in a low-traffic area near the work area and it is best to use disposable towels. An emergency eye-wash should be installed near the hand-washing sink and there should be access to an emergency shower in or near the laboratory.

The patient waiting room also needs special attention. Separate waiting areas and toilet facilities are advised for injected patients. The facilities shall include a washbasin as a normal measure of hygiene. Washrooms designated for use by nuclear medicine patients should be finished in materials that are easily decontaminated. Hospital staff should not use the patient washing facilities as it is likely that the floor, toilet seat, and tap handles will be contaminated frequently.

A large amount of activity is kept in a nuclear medicine department, potentially giving rise to a significant amount of exposure. However, if the radionuclides are stored in shielded containers and if shielded barriers are used when handling the activity in the laboratory, the exposure is kept below reasonable levels.

It is easier to arrange shielding of a source than to build in structural shielding. However, in addition to shielding of the individual sources, it may be necessary to incorporate some shielding into the handling area walls, particularly if the occupancy rate in adjacent locations is high. The patient, it must be recalled, is also a source of exposure and it must be investigated if structural shielding is needed in the waiting rooms and examinations rooms.

The exposure rates in the areas just outside the department or clinic must be investigated. In facilities using PET substances, the exposure becomes higher and it has been found that structural shielding is needed [06Mad]. The need for shielding depends on the location of the rooms used. Great care will be required in the positioning of the rooms in relation to public access areas.

Security of the sources is also very relevant in nuclear medicine. In addition to open sources, the clinic often possesses sealed sources used for quality control purposes. Access to the clinic must be restricted. Only laboratory staff should be permitted in the room when radionuclides are being handled, and the doors and windows should be lockable. A high security level is required in waste storage rooms, allowing only a few members of the personnel to have access to the area. A source storage area and an area for temporary storage of radioactive waste must be arranged. When the clinic is designed, access control

should also be considered for the source storage areas and for rooms where hospitalised patients are undergoing radionuclide therapy.

The objective of source security is to ensure continuity in the control and accountability of each source at all times. The department should have an inventory of sources received and it is important to develop procedures to ensure the safe movement of radioactive sources within the hospital at all times, from reception to disposal. It is also advisable to have a security system to prevent theft, loss, unauthorised use, or damage to sources, or entrance of unauthorised personnel to the controlled areas.

5.3.6.1.1 Special considerations concerning waste

A great amount of waste is produced in a nuclear medicine department. The solid waste includes cover papers, gloves, empty vials and syringes, radionuclide generators, items used by hospitalised patients after radionuclide therapy, sealed sources used for the calibration of instruments, and biological waste.

Depending on the final handling or disposal, the waste could be divided into different categories: (A) Waste that will end up in a public waste treatment system, with or without burning; (B) waste that will be discharged into the public sewage system; and (C) waste that will be disposed in a special plant for used radiation sources.

Solid waste could, after proper treatment and conditioning, be handled in the public waste treatment system. There are special transport regulations and also specific requirements from the plant accepting the waste. Discharge limits are given by the regulatory authority. Solid waste containing short-lived radionuclides could sometimes be stored for decay. Thereafter it may be handled in the public waste system if the discharge is within the limits given in regulations. Radionuclide generators could be stored for decay, checked for contamination, and dismantled or, preferably, returned to the producer.

Liquid waste includes residues of radionuclides, patient excreta, and liquid scintillation solutions. If properly treated and conditioned, some liquid waste could be handled in the public waste system. For organic and infectious liquid waste, burning is a suitable method. Treated aqueous effluents can be discharged into the environment via the sewage system if either clearance has been granted for the radioactive substance or the discharge is within the limits given in regulations. Sometimes liquid waste containing short-lived radionuclides can also be stored for decay in order to permit subsequent handling by the public waste system.

Containers to allow separation of different types of radioactive waste should be available in areas where the waste is generated. The containers must be suitable for their purpose (volume, shielding, leak-proofing, etc.) Each type of waste should be kept in separate containers, properly labelled to supply information about the radionuclide, physical form, activity, and external dose rate. For example, broken glass-ware, syringes, etc., must be placed in separate containers to prevent personnel being injured.

For diagnostic patients there is generally no need for collection of excreta and ordinary toilets can be used. For therapy patients, policies differ between countries. In some countries, a special delay tank has been installed in order to minimise the amount of activity before releasing it into the common sewer system. However, this technique has been found less efficient.

A room for interim storage of radioactive waste should be available. The room should be locked, properly marked, and if necessary ventilated. This room should be kept tidy and in good order. Flammable waste should be placed apart from other waste. It is essential that all waste be properly packed in order to avoid leakage during the storage. Biological waste should be refrigerated or put in a freezer. Records should be kept where the origin of the waste can be identified.

5.3.6.2 Optimisation of the protection of the patient

Radiation doses to the patients vary also in diagnostic nuclear medicine [05Har], for different examinations and for the same examination done in different clinics. One variation will be due to the use of different radiopharmaceuticals for the same examination. The justification of the different

examinations techniques used is an important task for every clinic. Likewise, if different amounts of activity are used for the same examination, this needs careful justification.

Information about the nominal amount of activity for the different examinations is often easy to collect. However it is usually considered acceptable that the actual activity administered to the patient differs by sometimes as much as 25 % from the nominal amount, which results in the same variation in radiation dose to the patients. Thus when assessing radiation dose to the patients the actual administered activity is best used in the study.

For many examinations, the amount of activity required depends on the size of the patient, but the administered activity, at least for adult patients, is not always adjusted accordingly. One reason for this is that it is difficult to place an exact amount of activity in the syringe. There are additional considerations for each individual patient. For example, the patients' physical condition influences how long they are able to lie on the examination table, and the radiation dose can be reduced if longer examination times are accepted, the longer time the less radiation dose is needed. Finding out how other practices are dealing with such factors could be useful.

Other measures, post treatment, can be taken to reduce radiation dose. These include hydration and frequent voiding of the urinary bladder, using laxatives, or introducing special food: fatty meal forces voiding of the gall-bladder, thus reducing the time that the pharmaceutical remains in the organ/body. The biological half-time of the radiopharmaceutical in the body is a very important factor influencing the radiation dose.

The typical radiation dose level used in a clinic is extremely difficult to evaluate unless there is something to compare with. Thus, diagnostic reference levels are useful also in nuclear medicine. [Chapter 3] The levels could be derived in a similar manner as for diagnostic radiology. A number of clinics regionally or nationally could contribute to the assessment of the range of radiation doses encountered.

The administered activity to the patient is a useful quantity in nuclear medicine. A diagnostic reference level could be derived from the distribution of resulting doses. In addition to the obvious information to be registered concerning the assessed activity, information on equipment used and acquisition time, etc., can be very useful in order to explain any differences in radiation dose to the patients.

When an organ of an individual patient is being examined or treated, its function and size will determine the radiation dose to that organ, and the activity given should be assessed individually. For example, the function of a diseased thyroid varies considerably, with a significant effect on uptake and dose. It is very important that any pregnancy is discovered prior to treatment. The radiation doses to the child to be born are much higher than for diagnostic procedures.

5.3.6.2.1 Special considerations: children

It is considered bad practice not to adjust the amount of activity administered to children. Several methods exist to adjust and determine the activity when children are examined. The body size differs very much between children of different ages, and thus the adjustment must be based on a size-dependent factor such as weight, body mass index, or body surface area. A suitable fraction of the activity used for adults could be derived in this way. However, there is no general rule which can be applied to all types of examination. A separate evaluation is required for each type of examination.

The adjustment should lead to an administration of activity resulting in the same counting statistics for all ages in order to achieve the same diagnostic accuracy. The European Association of Nuclear Medicine provides dose cards online (www.eanm.org). An additional complication that may need to be considered is that it can be difficult for children to be co-operative during the examination. Special attention to such problems is needed in the clinic; any decision will be based on the conditions for individual patient. Further reading see Chapter 3.

5.3.6.2.2 Special considerations: the pregnant patient

As with other medical procedures, it is important to know if a female patient is pregnant or not when an examination or therapeutic procedure is requested. Preferably, this information should be available before the patient has arrived at the clinic, so that the nuclear medicine department has time to consult the referring physician.

The clinic should consider postponing the examination until after the pregnancy. These decisions have to be taken on an individual basis. The clinic should be able to reduce the activity and allow longer acquisition times, or adjust the procedure in other ways in order to lower the dose. One example of the latter is to perform and evaluate perfusion scans before any ventilation examination, as the latter is unnecessary if the first is found normal. Post examination activities such as drinking water and frequent voiding of the bladder could also be considered.

The exact dose to the fetus is not very easy to assess, but generally it depends on the amount of activity used. The doses to a fetus from nuclear medicine examinations are usually low, in the same region as in diagnostic radiology. Nevertheless, the clinic should not neglect to inquire whether the patient might be pregnant, and optimisation of protection is important for all examinations. [Table 5.3.4](#) gives some examples of radiation dose to fetus, from examinations performed early in the pregnancy and all using ^{99m}Tc [[00ICR1](#)].

Table 5.3.4. Approximate radiation dose to the child to be born.

Nuclear medicine examination	Absorbed dose [mGy]
Renal DTPA	6-9
Bone scan	2-5
Lung perfusion (MAA)	0.4-0.6
Lung ventilation (aerosol)	0.1-0.3

Sometimes, post-treatment consideration is necessary. In some cases, it is advisable not to become pregnant for a period of time after a nuclear medicine examination due to a relatively long effective half-life of the radiopharmaceutical. This applies, for instance, to examinations using Se-75 or Fe-59. The patients must be duly informed in each specific case.

5.3.6.2.3 Special considerations: patients who are breast-feeding

Before administering a radiopharmaceutical to a woman who might be breast-feeding, the clinic should inquire whether this is the case. If the woman is indeed breast-feeding, it should be considered whether the examination could be postponed until after the mother has stopped breast-feeding. Terminating breast-feeding is recommended after many nuclear medicine procedures (further reading Chapter 4), but for some procedures a different pharmaceutical can be chosen in order to reduce the radiation dose to the child.

5.3.6.2.4 Special consideration: introducing PET/CT and other multi-modal examinations in the clinic

New techniques where different modalities are combined, such as PET/CT, are now common in the nuclear medicine department [[08IAE](#)]. The radiation dose for examinations also including a CT examination is of course higher than only using PET. How much is depending on what image quality is needed from the CT scan. If a high image quality is needed, comparable with an ordinary diagnostic examination, the total radiation dose can be significantly higher than for the PET scan alone. One review [[08ZA](#)] indicates that a FDG-PET examination could consist one third of the total radiation dose when a

CT examination with diagnostic quality is added the PET examination. But when a non-diagnostic quality CT is added the contribution from the PET is about 90% of the total radiation dose from the examination.

The total radiation dose is needed to be surveyed for these examinations.

The use of such multi-purpose machines actually emphasises the need for co-operation between experts in nuclear medicine and X-ray diagnostics. The optimisation of radiological protection must be performed simultaneously for both modalities and this could not be done without team-work. Education and training activities must be adapted as many of the staff are likely to be trained in one of the fields only, and it is important that all aspects of the technique is understood by the personnel performing the examinations.

5.3.6.2.5 Special consideration: carers and comforters

In therapeutic nuclear medicine some actions to optimise radiological protection of carers and comforters is needed. Two important factors that will influence the radiation dose to carers and comforters are of course time spent with the patient and the distance from the patient. Information on how the carers could minimise radiation dose have to be given to carers and comforters. Restrictions should be required.

Carers and comforters must also be aware that contamination may occur, especially in the toilet area, and that special hygiene measures such as careful washing of hands, etc., are required. The instruction should be presented both orally and in writing. Suggestions concerning the wording of such information are available [99Mou].

5.3.6.3 Protection of the personnel and general public

A variety of work activities will contribute to the level of external irradiation, amongst others handling of delivered radioactive sources, activity measurements, storage of sources, patient examinations, care of ‘radioactive’ patients, and handling of radioactive waste. The main contributions to internal irradiation are preparation of radiopharmaceuticals, administration of radiopharmaceuticals, and in some cases incidents in the laboratory or patient care. Each work activity has to be investigated and the work optimised in order to keep the radiation dose as low as reasonably achievable.

A programme of personal dose monitoring for all workers is recommended; this will include nuclear medicine physicians, nuclear medicine physicists, nuclear medicine technologists, nuclear medicine nurses, and radiopharmacists is needed. Personal monitoring for external exposure is almost always needed. Monitoring for internal exposure is also sometimes needed and possible, e.g. for personnel involved in the iodine therapies. A programme for area monitoring of the workplace, including both contamination monitoring and dose rate monitoring, is also necessary. Continuous monitoring with an area monitor should be considered for source storage and handling areas.

Doses to the hands can be substantial and have to be assessed. It is sometimes rather hard to monitor hand and finger doses routinely, but frequent measurements should be performed, especially when new techniques has been introduced or new personnel performs the work. Further advice is available [08ICR].

Nuclear medicine and radiopharmacy staff must use syringe shields and vial shields. Shielding is primarily used to protect against gamma radiation, but is also necessary to protect individuals against some high-energy beta-emitting radionuclides such as ^{32}P and ^{90}Y . The shielding material and thickness depend on the types and amounts of radionuclides used. The most common shielding material for gamma rays is lead, but high-energy beta emitters such as ^{32}P require an inner shield of a low atomic number material, e.g. perspex, to shield bremsstrahlung production.

Suitable personal protective equipment must be provided, used, and properly maintained. Protective clothing should always be used in working areas where there is a likelihood of contamination. In nuclear medicine, the function of the equipment is not just to reduce external doses but also to protect the person adequately if contamination occurs. Such protective clothing includes lab coats, gloves, etc. It is also important that personnel wash their hands frequently.

In some cases, when handling high-energy beta sources, e.g. ^{32}P , ^{89}Sr , ^{90}Y , or ^{188}Re , it is appropriate to wear *safety glasses*. This will reduce the external irradiation of the eye as well as prevent the high

radiation doses to the eye that may result from accidental contamination. Forceps or tongs should be available and used to reduce the radiation exposure by increasing the distance between the source and the hands. Shielded containers and syringe shields are important to reduce the external dose, especially to the hands.

Contaminations are likely to occur from time to time. Therefore, personnel must have both equipment to deal with spills (emergency kits) and training on how to act and start decontamination procedures [05Lan2] Equipment for ad hoc monitoring purposes should also be available in the department. On the basis of possible events identified by a safety assessment, emergency procedures should be trained. Such events include loss during shipment of sources, damage to sources (e.g., Tc-generators), radioactive spills, and medical emergencies. The procedures should be clear and posted visibly in places where their need is anticipated.

Diagnostic and therapy nuclear medicine procedures in general do not imply any great doses to the general public outside the department although as mentioned before the radiation environment has to be assessed outside the department. However, a risk assessment, primarily for carers, has to be carried out when patients are released after treatments and restriction is sometimes needed.

5.3.6.3.1 Special consideration: hospitalised patients during treatment

All personnel handling the patient should be properly informed about working procedure and what the risks are. In particular, they need to know how to treat situations such as vomiting patients or handling soaked linen [01Gre]. Movable shields or the use of a lead apron should be considered. Special attention to any comforters is needed also for this field.

After patient discharge, any contaminated waste should be removed following decontamination of the room and equipment, and final survey of the room should be performed and documented. When the decontamination procedures and surveys with satisfactory results are complete, the room is clear for general use.

5.3.6.3.2 Special consideration: considerations when patients leave the hospital after treatment

The activities administered for diagnostic purposes are moderate, and the patient does not normally need to be hospitalised. For almost all diagnostic procedures, the maximum dose that could be received by another person due to external exposure from the patient is a fraction of the annual public dose limit and it should not normally be necessary to issue any special radiation protection advice to the patient's family.

However, the management of patients undergoing radionuclide therapy has to be designed to minimise radiation exposure to other persons [04ICR2]. The precautions to be taken for ambulatory patients treated for thyrotoxicosis depends upon the amount of radioactivity administered, the radiation dose rate in the vicinity of the patient, and expected normal patterns of daily contact between the patient and others.

All patients should be advised on basic hygiene measures (toileting, hand-washing, etc.) to minimise contamination of their home and work environment. Females should avoid pregnancy for at least 6 months following therapy, in case follow-up examinations might indicate the need for further radiation therapy. In addition, breast-feeding mothers undergoing radioiodine therapies are to be advised that complete termination of breast-feeding is necessary (further reading Chapter 4).

5.3.6.3.3 Special considerations: death soon after treatment

Occasionally, the condition of a patient treated with radionuclides may lead to death while the body still contains substantial residual activity. The rules for embalming, burial, or autopsy of cadavers containing substantial residual activities vary from country to country since they take account of various social, climatic, and religious factors.

In the event of the death of a patient who has recently received a therapeutic dose of a radionuclide, care has to be taken to ensure that workers and members of the public receive as low a dose as reasonably achievable at all stages prior to the burial or cremation. A plan should be described in written instructions to the hospital staff if death occurs while the patient is still in the hospital.

When a patient is released from the hospital with significant amounts of radioactive material, instructions on how to act in case of the patient's death must be given to the care providers. The most important instruction should be to contact the nuclear medicine department as soon as possible, and to delay any decisions about proceeding with arrangements for autopsy or disposal of the corpse until expert advice has been obtained.

The department should be consulted immediately to determine (by direct measurement or calculation) the amount of residual radioactivity in a cadaver. If there is still significant residual radioactivity, the physician who declares the patient dead should attach a label to the body indicating the presence and amount of radioactivity and the radionuclide. The label should be readily recognisable, legible, and easy to understand. The precautions to be taken in handling such cadavers depend on the nature and quantity of the radionuclide present and on the type of handling intended (e.g., autopsy or merely simple treatments prior to burial). As a general rule it can be said that no appreciable hazard exists unless the body is opened, when the hands and face of the pathologist could receive high radiation doses, depending on the duration of exposure and the dose rate.

The autopsy of highly radioactive cadavers should be restricted to a minimum. It is essential that the staff wear disposable gloves, and supplementary measures for radiation protection and decontamination should be provided.

5.3.7 Therapeutic radiology

5.3.7.1 Designing the facilities and structural shielding – special consideration in radiation therapy

The machines and sources used in radiation therapy all need significant shielding when used, and in the case of sealed sources also when stored and transported. Planning of each facility and treatment room is needed in order to optimise the shielding [06IAE]. In this context, it is important to use realistic assumptions when calculating the required dimensions of the shielding. 'Conservative', i.e., unrealistically pessimistic, assumed occupancy factors, etc., will lead to sub-optimal and extremely costly shielding arrangements.

The location in the building will very much determine shielding requirements for a radiation therapy installation. Present and possible future activities around the prospected location must be examined in the planning process in order to reduce the shielding needs. The result of such planning is often a treatment room located at the ground floor or even beneath the ground level; this reduces the need for shielding in the floor. Sometimes no activities above treatment machines are planned and the access of the roof is limited in order to minimise the shielding in the ceiling of the facility. Future needs must also be considered in the planning process: work activities around the treatment rooms could change and new machines and sources could be introduced. Hence, proper planning for therapy activities takes a lot of time.

The main shielding material used is concrete (O6NCR), and sometimes iron ore is added to increase density and thus shielding properties. The literature gives attenuation properties for a number of materials but usually, exact properties of the material used are not available and sometimes, the exact energy distribution in the radiation field of the accelerator is also unavailable. This, together with the fact that usage and occupancy factors have to be assessed in advance, often generates a 'worst case' scenario that will result in sufficient shielding.

However, weak spots can occur, for example it is very difficult to construct doors with the same shielding properties as the walls. Therefore, in order to reduce the need for shielding and the radiation dose at the entrance to the room, a so-called maze is included: a passageway that will result in multiple

scattering of the x rays and thus reduction of the energy. If this reduction is sufficient, there may even be no need at all of a shielded door.

The features of the equipment, as mentioned before, will define how much shielding is needed. In a linear accelerator, neutrons will be produced if the energy of the X-ray spectrum exceeds about 10 MeV [07Rud]. These neutrons can react with different parts of the accelerator, the treatment couch, and the patient, and induce radioactivity for example aluminium or copper. Air is also activated but if the room is properly ventilated this will not be of great concern. The half-life of the produced nuclides ranges from seconds to days, but during treatment this will not be of great concern. However, during physics measurements, service, and maintenance, the accelerator could be running for a longer time and therefore entering the room may have to be delayed in order to avoid unnecessary irradiation. If inner parts of the accelerator are to be removed during maintenances it must be acknowledged that the parts could be radioactive.

Special attention must be paid to the design of treatment rooms for therapy in order to reduce the risk of accidents and unnecessary exposure of personnel, patients, and comforters. From the control room, the personnel should have a view of the entrance of the treatment room and access ways such as corridors and other surrounding areas. The personnel in charge of the treatment must be able to view the patient on the treatment couch in order to be able to stop the treatment if the patient moves. This must be taken into account in the planning of the facility.

Surveillance cameras could be mounted so that a view of the whole treatment room is possible, and with an intercom system it is possible to communicate with the patient. This is especially important for low dose-rate (LDR) units where the patient is treated for a long time, and when possible the nurses could comfort the patient from afar.

5.3.7.1.1 Special considerations: accident prevention

Warning signs and lights should alert persons that a treatment is performed; sometimes an audible warning is also in place. Emergency buttons should also be installed. These should be placed in multiple places, of course also in the control room.

If possible, interlocks should be installed in doors and other entrances into the treatment rooms; this can easily be done in treatment rooms for accelerators. Such interlocks can also be considered for brachytherapy rooms, but sometimes they cannot be installed if the treatment is done with manual loading devices. Inside or outside brachytherapy rooms, radiation monitors have to be placed in order to indicate that the source has returned into the unit and that no source remains inside the patient when removed from the room.

New treatment types are introduced and new technology is also introduced like in many activities in the hospital as mentioned before, and radiation therapy is one of the high-risk activities where patient safety must be given high priority. One of many important stages in the treatment is dose planning and converting data into the simulation stage and finally to the treatment machine. In all these data transfers there is a risk that a faulty value is inserted.

Certain measures have to be taken in the daily routine in the hospital. One such issue could sound very basic but is important, viz., the need for checking the identity of the patient. There could be vast consequences if the wrong patient is treated.

The dose given to the patient could be verified with so-called in-vivo measurements where dose to the patient is measured and compared with calculated dose for this position. These measurements should be performed when there is still time to adjust the treatment regime.

Verifying the volume irradiated and thus the positioning of the patient is also important. With the availability of other equipment than film, the verification could be performed more frequently if this is deemed necessary.

5.3.7.2 Optimisation of the protection of the patient

The protection of the patient is an integrated part of dose planning. Organs at risk should be identified, but also the probability of complications in other tissues has to be considered. In many cases it is not difficult to kill the cancer cells but the difficulty is to reduce the side effects to an acceptable level [08Pur]. The prerequisite is careful diagnostics, individual planning of the treatment, and verification during treatment.

In order to acknowledge late effects of the irradiation, the patients have to be followed up not only during the treatment period but also after treatment. During the period of treatment physicians and nurses with special knowledge of radiation attend the patient, but after that when the patient is in less contact with the radiotherapy department, effects perhaps are not recognised as connected to the radiation dose received.

The follow-up serves several purposes, most importantly of course that the patient should receive proper treatment for complications, but also that the method and treatment regime are evaluated so that the radiotherapy department gets feed-back on specific treatment regimes. Thus physical complications, such as reduced lung capacity, should also be evaluated. The follow-up should be systematic in order to be part of the evaluation of the treatment. Minimising late effects is of particular importance for cancer types with a high expected survival time where the patient can be assumed to live long after the treatment with an expectation of having a certain life quality.

In this context, the risk of secondary cancer induction is also important and this should be included in the evaluation. However, secondary cancer is manifested many decades later which emphasises the need for assessing the risk when setting up the treatment. The risk estimates must be investigated for specific groups of patients since the sensitivity to radiation differs between different patients, and the dose distribution will vary between patients.

Justification of the treatment regime is important in radiotherapy too. For many cancers, different treatment regimes could be suggested, including surgery and medication. Sometimes, a high risk of late effects must be accepted, if no other treatment is possible or the treatment result for the particular patient is insufficient.

5.3.7.2.1 Special considerations: diagnostic procedures

A large amount of diagnostic procedures is carried out before, during, and after treatment [07Whi]. There is an evident danger that radiation protection is neglected during these procedures, since one may assume that the radiation treatment itself gives the bulk of the dose to the patient.

However, it must be kept in mind that doses outside the treated volume are modulated during treatment while diagnostic procedures irradiating a large volume, such as a computed tomography examination, will give a rather homogenous dose in the exposed volume. If one diagnostic procedure is performed for every treatment, the total dose due to diagnostics will be significant. Thus, optimisation of radiological protection in diagnostic examinations remains important even during radiotherapy with its, seemingly, very much bigger doses.

5.3.7.3 Protection of the personnel and general public

Typically, occupational doses are rather low in radiation therapy [08UNS]. Most treatments can be conducted with a minimum or no radiation dose to the personnel. However, if an accident happens, the radiation dose could be significant. Therefore, considerable efforts should be spent in order to prevent accidents.

Interlocks in doors and limited access to premises will reduce the risk of accidentally irradiating personnel and members of the public. However, occasionally people will be left in the treatment room by mistake. This could be both hospital personnel and people from outside such as maintenance staff. These

situations often demand the assessment of radiation dose and it is helpful if the person(s) concerned have worn a dosimeter.

Work procedures involving the handling of sources have become less frequent because after-loading techniques used in brachytherapy reduces the amount of manual handling. Nevertheless, situations do occur when personnel have to handle sources. When permanent seeds are placed in the patient's body, the personnel performing this have to be trained and routines have to be developed in order to have control over the sources used.

The personnel must be trained to handle acute situations, such as sources not being secured in after-loading brachytherapy. Equipment must be placed outside the treatment room, such as tweezers, tongs, shielded storage for the radiation source, monitor equipment, etc. Furthermore, the procedures must be practiced. One important decision that needs to be taken is whether pregnant personnel should be part of the treatment team, including medical doctors as well as emergency personnel recruited if the patient needs medical care in an acute situation. Any decision has to be communicated in the hospital.

5.3.7.3.1 Special considerations: discharge of patients after treatment with permanent implants

Radiation protection issues arise for patients treated with permanent implants who are discharged from the hospital [05ICR]. While there are no major difficulties, the patient must be informed about specific precautions just after the release from the hospital. The information to the patient must be provided in writing, easy to understand, in a language the patient is familiar with, and although written down it also has to be discussed with the patient.

The decision to discharge the patient should be based on dose constraints to the public, taking account of both the dose rate around the patient and how the patient lives. In some cases, the patient's personal situation may preclude an early discharge. The considerations will be similar to those that are required in nuclear medicine therapy.

To avoid radiological consequences to other persons in case of death or emergencies, the patient should carry information at all times about the treatment and whom to contact. If permanently implanted seeds are expelled or lost in such circumstances, the clinic and the appropriate authorities must be informed so that suitable action can be taken.

The external radiation level from a patient with permanent implants is not very high, so restrictions are only recommended for children and pregnant women. If the patient's partner is a pregnant woman, measurements could be performed after the implantation in order to assess whether any distance restriction is needed to ensure that the dose to the child to be born will not exceed 1 mGy for the remainder of the pregnancy. Whether such restrictions are necessary will depend, for instance, on the patient's weight.

The patient must be told how to take care of any seeds passed through the urine. During the first couple of days it is advisable that the patient passes urine through a sieve, and that the hospital provides a kit of a tweezers and a small container. If a seed is released, it is placed in the container and returned to the hospital. In instruction it should also be indicated how long these precautions are needed, this will depend on radionuclide used. Precautions are also advisable when having sexual intercourse. The use of condoms is recommended for a couple of weeks after the implantation. The patient should also be told that there may be an added risk, albeit limited, of genetic effects to the child.

Information about diagnostic and therapeutic procedures performed on the patients must be documented and stored. If surgical procedures are performed, the hospital needs to be aware of the seeds and remove any seeds found with tweezers or similar. This information is also needed if the patient dies and the body is about to be cremated. Various precautions are recommended in order to protect the staff in the crematorium and the environment [05ICR].

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5.4 Magnetic Resonance Imaging, Diagnostic Ultrasound

Medical Radiological Protection

J.H. BERNHARDT

5.4.1 Introduction

Magnetic resonance imaging (MRI) has become an established diagnostic imaging modality. The clinical usefulness of in-vivo magnetic resonance spectroscopy (MRS) has been demonstrated in several clinical applications and is being explored further. It can image soft tissues – unobstructed by bone – with enhanced contrast. Moreover, the ability to provide images in numerous planes without requiring the repositioning of the patient has rendered MRI a very effective and important tool for soft issue imaging. These techniques involve exposure of the patient to static and time-varying magnetic fields, radiofrequency electromagnetic fields, and acoustic noise. In particular exposure conditions, these fields may pose a health hazard or increased risk.

The MR technique will be explained in the corresponding sections of LB Vol. VIII/7B.

In Sects. 5.4.2.1 and 5.4.2.2 below the protection of the patient, the personnel and the general public is discussed.

Diagnostic ultrasound has been widely used in clinical medicine for many years with no proven deleterious effects. However, if used imprudently, diagnostic ultrasound is capable of producing harmful effects. The range of clinical applications is becoming wider, the number of patients undergoing ultrasound examinations is increasing and new techniques with higher acoustic output levels are being introduced. It is therefore essential to maintain vigilance to ensure the continued safe use of ultrasound.

The ultrasound technique will be explained in the corresponding sections of LB Vol. VIII/7B.

In Section 5.4.3.1 below the protection of the patient is discussed.

5.4.2 Magnetic resonance imaging

Except acoustic noise, three types of magnetic fields are employed in MR imaging and spectroscopy:

- a high static magnetic field generating a macroscopic nuclear magnetisation;
- rapidly alternating magnetic gradient fields for spatial encoding of the MR signal; and
- RF electromagnetic fields for excitation and preparation of the spin system.

In particular exposure conditions, these fields may pose a health hazard or increased risk for the patient. The biophysical interaction mechanisms of these fields are discussed in Sect. 2.2.2.3; and biological effects in Sect. 2.3.2.3.1. Several reviews concerning safety aspects of clinical MR procedures have been published (e.g. [00Ord1, 01She1, 03She1]. These publications, in conjunction with other reviews and recent literature, form the basis for the review on MRI procedures and its safety aspects in the sections below.

5.4.2.1 Protection of patients and volunteers undergoing MR procedures

Clinical experience currently indicates that adequate diagnostic information can be obtained while examining a patient for periods ranging from 5 min to over 1 hour and may be repeated several times over the course of the disease or abnormal condition. In some cases, due to the use of MR spectroscopy or interventional MR procedures, examinations can last several hours.

The following section provides a brief summary of exposure limits and precautions to be taken in order to minimize health hazards and risks to patients and volunteers undergoing MR procedures according to

- the technical product standard IEC 60601-2-33 issued by the International Electrotechnical Commission in 2001 [01IEC1]; and
- the safety recommendation issued by the International Commission on Non-Ionizing Radiation Protection [04ICN1, 09ICN2].

Because of the uncertainty over identified deleterious effects, it is recommended that exposure limits be divided into three tiers:

- routine MR examinations for all patients (normal operating mode);
- specific MR examinations outside the normal operating range where discomfort and/or adverse effects for some patients may occur. A clinical decision must be taken to balance such effects against foreseen benefits; exposure must be carried out under medical supervision (controlled operating mode);
- experimental MR procedures, at levels outside the controlled operating range, for which special ethical approval is required in view of the potential risks (experimental operating mode).

Static magnetic fields

Until now, most MRI or MRS examinations have been made using static magnetic fields up to and including 3 T, although whole-body MR systems with static magnetic fields up to 8 T are already used in clinical tests. Higher magnetic flux densities offer potential diagnostic advantages, particularly for MRS. The literature does not indicate any serious adverse health effects from the exposure of healthy human subjects up to 8 T. However, it should be noted that, to date, there have been no epidemiological studies performed to assess possible long-term health effects in patients or volunteers. The greatest potential hazard comes from metallic, in particular ferromagnetic materials (such as scissors, coins, pins, oxygen cylinders) that are accelerated in the inhomogeneous magnetic field in the periphery of an MR system and quickly become dangerous projectiles. The risk can only be minimized by a strict and careful management of both patients and staff.

Several studies have reported that individuals exposed to static magnetic fields above 2-3 T experience transient sensory effects associated with motion in a static field gradient such as nausea, vertigo, a metallic taste, and magnetic phosphenes when moving the eyes or head [92Sch1, 06deV1, 06deV2, 07Atk1, 07Glo1]. Therefore it is recommended that examinations above this static magnetic flux density be conducted under especially careful medical supervision. The occurrence and severity of these symptoms can be decreased by slowing the rate of motion of an individual through the magnetic field gradient [06Cha1].

The recommended upper limit for clinical routine whole-body exposure to static magnetic fields is 4 T (normal operating mode), due to the limited information concerning possible effects above this static field strength [02IEC1, 04ICN1, 09ICN2]. At higher field strengths applications should be made based on critical risk/benefit analyses and informed consent of the subject. Additionally, more careful monitoring of subjects is required, although physiological monitoring in an MR scanner does itself increase the risk of RF burns, which is however an avoidable risk.

Current information does not indicate any serious health effects resulting from exposure to static magnetic fields up to 8 T. It should be noted, however, that such exposures can lead to potentially unpleasant sensory effects such as vertigo during head or body movement. For the controlled operating mode, there should be an upper limit for whole-body exposure of 8 T [09ICN2].

The FDA deems magnetic resonance devices to be a significant risk when the main static magnetic field is greater than 8 T for adults, children, and infants aged >1 month, and greater than 4 T for neonates, i.e. infants aged 1 month or less [03FDA1].

Time-varying magnetic gradient fields

The rapidly switched magnetic gradient fields used in MRI for spatial encoding induce electric fields in the human body in accordance with Faraday's law which, if of sufficient magnitude, can produce nerve

and muscle stimulation (see Sect. 2.3.2.4.2). The induced electric field is proportional to dB/dt , the time rate of change of the magnetic field. From a safety standpoint, the primary concern with regard to time-varying magnetic fields is cardiac fibrillation, because it is a life-threatening condition. In contrast, peripheral nerve stimulation is of practical concern because uncomfortable or intolerable stimulations would interfere with the examination (e.g., patients movements) or would even result in a termination of the examination.

The threshold for cardiac stimulation is well above the level for intolerable stimulation, except at very long pulse durations which are not clinically useful. The trend in clinical MRT is toward shorter gradient pulses to reduce imaging time.

The study on nerve stimulation [01Nye1] is the largest study to date, and because it distinguishes between varying levels of stimulation (i.e., threshold, uncomfortable, intolerable) it provides a basis for recommended exposure levels for time-varying magnetic fields. Many medical procedures are uncomfortable or painful, but are tolerated by patients for the medical benefit. However, intolerable stimulation would interfere with an examination and the patient would receive no benefit. Consequently, this is to be avoided. Bourland's data [99Bou1] indicate that the lowest percentile for intolerable stimulation is approximately 20% above the median threshold for peripheral nerve stimulation. Consequently, it is recommended that the maximum exposure level for time-varying magnetic fields be set equal to a dB/dt of 80% of the median perception threshold for normal operation, and 100% of the median for controlled operation. This median perception threshold is described by the following equation, which is an empirical description:

$$dB/dt = 20 (1 + 0.36/\tau) \text{ (in T/s)} \quad (5.1)$$

where τ is the effective stimulus duration in ms. The effective stimulus duration is the duration of the period of monotonic increasing or decreasing gradient. It is defined as the ratio of the peak-to-peak variation in B and the maximum value of the time derivative of B during that period.

Radiofrequency fields

Time varying electromagnetic fields with frequencies above 10 MHz that are used in MR procedures to excite and prepare the spin system, deposit energy in the human body that is mainly converted to heat. The parameter relevant for the evaluation of biological effects of RF fields is the increase in tissue temperature, which is dependent not only on localised power absorption and the duration of RF exposure, but also on heat transfer and the activation of thermoregulatory mechanisms leading to thermal equalisation within the body.

For whole-body exposures, no adverse health effects are expected if the increase in body core temperature does not exceed 1°C. In the case of infants, pregnant women, and persons with cardiocirculatory impairment, it is desirable to limit body core temperature increases to 0.5°C. Similarly, local temperature under exposure to the head, trunk, and/or extremities should also be limited to the values given in Table 5.4.1.

Since temperature changes in the various organs and tissues of the body during an MR procedure are difficult to measure in clinical routine, RF exposure of patients is usually characterized by means of the "specific absorption rate" (SAR in W kg^{-1}). As only parts of the patient's body are exposed simultaneously during an MR procedure, not only the whole-body SAR but also partial-body SARs for the head, the trunk and the extremities should be estimated on the basis of suitable patient models (e.g., [01Bri1]. Based on the published experimental studies concerning temperature rise and theoretical simulations, the SAR levels summarized in Table 5.4.2 should not be exceeded in order to limit temperature rise to the values given in Table 5.4.1. With respect to the application of the SAR levels defined in Table 5.4.2, the following points should be taken into account:

- Partial-body SARs scale dynamically with the ratio r between the patient mass exposed and the total patient mass. For $r \rightarrow 1$ they converge against the corresponding whole-body values, for $r \rightarrow 0$ against the localized SAR level of 10 W kg^{-1} defined for occupational exposure of head and trunk [98ICN1].

- The recommended SAR limits do not relate to an individual MR sequence, but rather to running SAR averages computed over each 6-min-period, which is assumed to be a typical thermal equilibration time of smaller masses of tissue [02Bri1].

Table 5.4.1. Basic restrictions for body temperature rise and partial-body temperatures.

Operating mode	Rise of body core temperature [°C]	Spatially localized temperature limits (°C)		
		Head	Trunk	Extremities
Normal	0.5	38	39	40
Controlled	1	38	39	40
Restricted	>1	>38	>39	>40

Table 5.4.2. SAR levels valid at environmental temperatures below 24°C, average time 6 min. (From [04ICN1]).

Operating mode	SAR (average time: 6min) [W kg ⁻¹]					
	Whole body	Partial-body		Local (average over 10 g tissue)		
		Any region except head	Head	Head	Trunk	Extremities
Normal	2	2-10 ^a	3	10 ^b	10	20
Controlled	4	2-10 ^a	3	10 ^b	10	20
Restricted	> 4	> (2-10 ^a)	> 3	10 ^b	10	> 20
Short term SAR	The SAR limit over any 10 s period should not exceed 3 times the corresponding average SAR limit					

^a) Partial-body SARs scale dynamically with the ratio r between the patient mass exposed and the total patient mass:

$$\begin{aligned} \text{-- normal operating mode: } \text{SAR} &= (10 - 8r) \text{ W kg}^{-1} \\ \text{-- controlled operating mode: } \text{SAR} &= (10 - 6r) \text{ W kg}^{-1} \end{aligned}$$

The exposed patient mass and the actual SAR levels are calculated by the SAR monitor implemented in the MR system for each sequence and compared to the SAR limits.

^b) In cases where the eye is in the field of a small local coil used for RF transmission, care should be taken to ensure that the temperature rise is limited to 1°C.

Acoustic noise

Generally, the acoustic noise produced during the MR procedures represents a potential risk to patients undergoing examinations on MR systems operating above 0.5 Tesla [00McJ1]. Internationally recommended limits for acoustic noise produced during MR procedures are based on recommendations for occupational exposures that are inherently chronic. There are no recommendations for non-occupational exposure to relatively short term noise produced by medical devices. For the safe use of medical equipment the acoustic noise must be restricted. Technical standards recommend that hearing protection shall be required for the safety of the patient if the maximum A-weighted r.m.s. sound pressure level of the MR equipment can exceed 99 dBA [01IEC1]. However, this noise level may not be appropriate for individuals with underlying health problems, who may have problems with noise at certain levels or at particular frequencies. Other guidelines recommend to limit the noise level at the ear of the patient for exposure times up to 2 h to 91 dBA [03SSK1].

It is recommended to offer hearing protection to the patients, when a noise level of 80 dB_A is exceeded. Hearing protection should always be worn by patients undergoing MR procedures at levels exceeding 80 dB_A, at best by headphones allowing verbal communication.

The exposure of staff and other health workers near the MR system is also a concern. This includes particularly those persons who are involved in interventional procedures or who remain in the room for

patient management reasons [00McJ1]. Several national guidelines recommend that hearing protection be worn by staff exposed to an average of 85 dBA over an 8-h day [03SSK1]. In view of the inherently chronic nature of occupational exposure and the comparably small inconvenience, it is recommended that staff follow the same rules for wearing hearing protection that is recommended for patients.

Because of the different problems mentioned above, ICNIRP strongly recommends the design of "quiet" gradient coils. Gradient coils windings can be designed such that all Lorentz forces generated by the pulsing of current are balanced [01Man1]. Greater coil stiffness and damping the coil should reduce mechanical vibration and associated noise.

Pregnant patients

There is at present insufficient knowledge to establish unequivocal guidance for the use of MRI procedures on pregnant patients. In these circumstances, ICNIRP recommends that MR procedures may be used for pregnant patients only after critical risk/benefit analysis, in particular in the first trimester, to investigate important clinical problems or to manage potential complications for the patient or fetus. The procedure should be conducted using a verbal and written informed consent procedure. The pregnant patient should be informed on the potential risks, also compared with those of other alternatives. Excessive heating is a potential teratogen; because of uncertainties in the RF dosimetry during pregnancy, it is recommended that exposure duration should be reduced to the minimum and that only the normal operation level is used. In addition, large doses of MRI gadolinium-based contrast agents have been shown to cause post-implantation fetal loss, retarded development, increased locomotive activity, and skeletal and visceral abnormalities in experimental animals. Such agents should only be used during pregnancy if the potential benefit justifies the risk to the fetus.

5.4.2.1.1 Contraindications and further considerations

Contraindications

Examinations of patients who have electrically, magnetically, or mechanically activated implants (e.g., cardiac pacemakers), or who rely on electrically, magnetically, or mechanically activated life-support systems, may be contraindicated for certain devices. Examinations of patients with ferromagnetic aneurism clips or certain metallic implants are also contraindicated. Lists of implants and materials tested for safety or compatibility in association with MR systems have been established and identified. This information is readily available on-line at www.MRIssafety.com.

Considerations related to patient's condition

Communication with the patient and/or monitoring (e.g., of the anesthetized patient) must be assured throughout the examination.

Certain patients may experience claustrophobia. The possibility of claustrophobic reactions should be explored before an examination is undertaken.

Because some individuals may exhibit an increased sensitivity to heating, it is advisable to ascertain if the patient's history comprises incidents indicative of sensitivity to heat and, if necessary, limit the duration of the examination.

Special requirements apply to the equipment and methods used for the monitoring of the patient under MR exposure conditions. Cardiorespiratory function may be monitored using non-ferromagnetic transducers to register the heart beat rate, blood pressure, and respiratory rate. The ECG may be subject to distortion because of electrohydrodynamic interactions and may not yield useful information. Non-perturbing fiber-optic probes for measurement of body temperature are available. Under MR exposure conditions, oral temperature and the temperature of the skin of exposed body parts are suitable for patient monitoring. Detailed information on body temperature measurements may be found in a review by Shellock [01She2].

Ferromagnetic objects are attracted by magnetic fields. Depending on size, composition, and location of metallic implants or inclusions, serious injuries may result because of motions and displacement of such objects. Moreover, the presence of such objects results in artefacts in diagnostic information. The presence or absence of such objects in the body has to be ascertained and the consequences evaluated

before an examination is undertaken. Additionally, adjustments with regard to the MR safety information may be required in consideration of the field strength of the MR system that is being used for the patient. For example, magnetic field interactions are different using a 0.2-T vs. a 3.0-T MR system [02She1].

Projectile/missile effects

The magnetic fringe field near the MR system's magnet may be strong enough to attract ferromagnetic objects and to cause them to fly towards the magnet. Thus, metallic objects, particularly with sharp edges, may become dangerous projectiles. All such objects have to be eliminated from the examination room and proper danger or warning signs must be posted. Details are presented in the IEC standard [01IEC1]. Guidelines to prevent projectile or missile effect accidents have been presented [03She1].

The manufacturer should provide information regarding the extent of the zone in which collision hazard and danger of uncontrolled movements of objects exist (e.g., uncontrolled movement of hospital carts, trolleys, loose tools, medical instruments, etc.).

Record keeping and patient follow-up

Examination records should be kept and patients should be monitored according to standard requirements of good medical practice. Observations on adverse reactions should be collected, reported according to national requirements, and published in the medical literature.

Electromagnetic interference

There are numerous reports of failures of electronic and mechanical medical devices due to electromagnetic interference (EMI) generated by the intense pulsed and static magnetic fields emitted by MRI systems. EMI involves the induction of deleterious voltages and currents into the circuitry of medical electronics, causing temporary or permanent failures of critical components. Damage of mechanical components due to forces induced by static magnetic fields also can be considered EMI. Non-fatal problems were experienced by several patients during MRI treatment involving each of the following devices: totally implanted and external infusion pumps delivering large drug overdoses, pulse oximeters for cardiac monitoring failing or providing erroneous data, and a ventilator malfunctioning during an MRI examination.

In addition, more than several dozen reports of burns exist for patients who were connected to external monitoring devices while receiving MRI imaging procedures. Most of these problems occurred when using ECG electrodes mounted on the chest, or pulse oximeter sensors on the finger tip. Heating is caused by RF power that is coupled into the metallic leads that connect an external monitoring instrument to objects mounted on the surface of the body.

The use of "MRI compatible" devices can prevent associated patient burns and the electronic and mechanical failure of such devices. MRI compatible devices are being designed using appropriate materials and wires or fiber optics to minimize coupling of the RF fields to medical devices.

5.4.2.2 Protection of the Personnel and General Public

In a conventional MRI system operating at 1 T, because of its design, it is unlikely that radiological staff would be exposed to significant fields. Some newer open 0.7 T MRI systems allow medical personnel to perform interventional procedures on patients under MRI guidance. It is possible that their hands, heads or torsos may receive significant exposure under such conditions, especially for gradient fields [04ICN1, 08ICN1]. The gradient field is lower than the static magnetic field but it is pulsed rapidly in time and is a function of imaging technique and design of the MRI system. It is significant to note that the time rate of change of the gradient magnetic field is closely related to the strength of the electric field induced inside the body.

The demand for increased spatial resolution and high signal-to-noise ratio (SNR) from MRI instruments has prompted the use of much higher static magnetic fields (as high as 11 T). This development has led to the use of higher RF frequencies for MRI, which, in principle, not only can augment the amount of RF power deposition inside the patient's body, but also increases the EMF exposure for workers using MRI equipment in the hospital environment and workers employed for

supporting, servicing, developing and manufacturing this equipment. There has been particular interest in the exposure of the head, torso, and limbs to the gradient fields, which may be substantial under certain operational environments.

In laboratory studies of humans, no pronounced effects on physiological parameters have been found from exposure to fields up to 8 T, except for a small increase in systolic blood pressure. Based on modelling, a clinically significant blood flow reduction is predicted only at field levels over 15 T. There is no evidence of effects of exposures up to 8 T on other aspects of cardiovascular function, or on body temperature, memory, speech or auditory-motor reaction time, or of any serious health effects in human volunteers. There is some evidence for effects on eye-hand co-ordination and visual contrast sensitivity. Fields of 2-3 T can cause transient sensory effects including nausea, vertigo, metallic taste, and phosphenes when moving the eyes or head; sensitivity varies between individuals, and the effects can be minimised or abolished by moving more slowly through the field [06WHO1, 07Van1, [09ICN1](#)].

There are few epidemiological data on long-term health in persons exposed to static fields, and none on potentially high exposure groups such as MRI operators. The available studies, on workers exposed up to several tens of mT in work in aluminium smelters, chloralkali plants, or as welders have had methodological limitations, but do not indicate strong effects from exposure of the above levels on cancer incidence, reproductive outcomes, or the other outcomes studied [06WHO1].

Based on review of the scientific evidence, ICNIRP recommends the following limits for exposure [[09ICN1](#)]:

5.4.2.2.1 Occupational exposures

It is recommended that occupational exposure of the head and trunk should not exceed a spatial peak magnetic flux density of 2 T. However, for specific work applications, exposure up to 8 T can be permitted, if the environment is controlled and appropriate work practices are implemented to control movement-induced effects. Sensory effects due to movement in the field can be avoided by complying with basic restrictions set in the ELF guidelines. When restricted to the limbs, maximum exposures of up to 8 T are acceptable.

5.4.2.2.2 General public exposures

Acute exposure of the general public should not exceed 400 mT (any part of the body), reflecting a reduction factor of 5 with respect to the occupational limits. However, because of potential indirect adverse effects, ICNIRP recognizes that practical policies need to be implemented to prevent inadvertent harmful exposure of people with implanted electronic medical devices and implants containing ferromagnetic materials, and injuries due to flying ferromagnetic objects, and these considerations can lead to much lower restriction levels, such as 0.5 mT [01IEC1]. The exposure limits to be set with regard to these non-biological effects are not, however, the remit of ICNIRP.

The rationale for these guidelines limits can be found in full in [[09ICN1](#)].

Table 5.4.3 summarises the limits.

Table 5.4.3. Limits of exposure^a to static magnetic fields. From [09ICN1].

Exposure characteristics	Magnetic flux density
Occupational ^b	
Exposure of head and trunk	2 T
Exposure of limbs ^c	8 T
General public ^d	
Exposure of any part of the body	400 mT

^a) ICNIRP recommends that these limits should be viewed operationally as spatial peak exposure limits.

^b) For specific work applications, exposure up to 8 T can be justified, if the environment is controlled and appropriate work practices are implemented to control movement-induced effects.

^c) Not enough information is available on which to base exposure limits above 8 T.

^d) Because of potential indirect adverse effects, ICNIRP recognizes that practical policies need to be implemented to prevent inadvertent harmful exposure of persons with implanted electronic medical devices and implants containing ferromagnetic material, and dangers from flying objects, which can lead to much lower restriction levels such as 0.5 mT.

5.4.2.2.3 Protective measures

ICNIRP recommends that the use of these guidelines should be accompanied by appropriate protective measures. These measures need to be considered separately for public places, where exposures to static magnetic field are likely to be very low and infrequent, and workplaces, where in some work situations strong static fields may be encountered. For members of the public, there is a need to protect people with implanted medical devices against possible interference and against forces on implants containing ferromagnetic material. In addition, in some specific situations, there is a risk from flying ferromagnetic objects such as tools. In work situations involving exposure to very high fields, there is a need for a set of site-specific work procedures intended to minimise the impact of transient symptoms such as vertigo and nausea. Further details can be found in [09ICN1].

5.4.3 Diagnostic Ultrasound

Diagnostic ultrasound has been widely used in clinical medicine for many years with no proven deleterious effects. The range of clinical applications is becoming wider, the number of patients undergoing ultrasound examinations is increasing and new techniques with higher acoustic output levels are being introduced. In particular exposure conditions, diagnostic ultrasound may pose a health hazard or increased risk for the patient.

The biophysical interaction mechanisms of ultrasound are discussed in Sect. 2.2.2.6, and biological effects in Sect. 2.3.2.2.2. Several reviews concerning safety aspects of clinical ultrasound procedures have been published (e.g. [97Bar1, 01Nyb1]). These publications, in conjunction with other reviews and recent literature, form the basis for the review of the safety aspects of diagnostic ultrasound in the sections below.

5.4.3.1 Protection of the Patient

5.4.3.1.1 Historical perspectives

The Bioeffects Committee of the American Institute of Ultrasound in Medicine (AIUM) was the first to publish authoritative statements on bioeffects and safety. In 1976 the AIUM published the conclusions of

a review of the early scientific literature. The specific wording has been carefully modified in succeeding years, but its essence is that there was no evidence of independently confirmed significant adverse biological effects in mammalian tissue exposed *in vivo* to intensities below 100 mW cm^{-2} (spatial peak temporal average intensity, I_{SPTA}) [91AIU1]. The situation has changed in the 1990s with the reports of lung capillary bleeding in animals following exposure to diagnostic levels of ultrasound [90Chi1].

In 1976, another significant event involved the inaugural meeting of the World Federation for Ultrasound in Medicine and Biology (WFUMB). The WFUMB has since evolved into a large international organization. The WFUMB has demonstrated a serious commitment to establishing conclusions and recommendations on safety based on international scientific consensus.

In the 1990s, has it been seriously considered that diagnostic applications of ultrasound might produce significant biological effects. Major ultrasound societies and organizations have paid attention to the safety of diagnostic ultrasound and progress has been achieved towards identifying international consensus on important issues. Knowledge on biophysical effects of ultrasound has become a fundamental consideration in the process of setting international standards for safety of ultrasound in medicine, an important function of the International Electrotechnical Commission (IEC).

In 1993 the AIUM in conjunction with the National Electrical Manufacturers Association (NEMA) developed a voluntary scheme for on-screen labelling of diagnostic ultrasound devices, known as the Output Display Standard [92AIU1].

5.4.3.1.2 Equipment output displays

The FDA Center for Devices and Radiological Health agreed to incorporate the AIUM/NEMA Output Display Standard (ODS) as a part of the FDA's pre-market approval process. These indicators are expressed in terms which reflect the potential for thermal and non-thermal biological effects. The EDA has maintained an overall maximum output limit of 720 mW cm^{-2} for all equipment [93FDA1]. Table 5.4.4 illustrates the increased available ultrasound exposure in terms of spatial peak temporal average intensity for equipment using the conventional application-specific FDA limits compared with the track 3 Output Display option for current, high-output devices.

Table 5.4.4. Comparison of maximum allowable spatial peak temporal average intensity (I_{SPTA}) limits for FDA-regulated application-specific and Output Display Standard (ODS) options.

	I_{SPTA} limit [mW cm^{-2}]	ODS track 3 option [mW cm^{-2}]
Peripheral vessel	720	720
Cardiac	430	720
Fetal, neonatal	94	720
Ophthalmic	17	720

High-output devices can be approved by the FDA for use in the USA with a maximum I_{SPTA} output of 720 mW cm^{-2} for all applications, as long as a thermal index (TI) and mechanical index (MI) are displayed for every possible setting of transducer type, output setting, focus, frame rate and pulse rate. Consequently, for fetal applications the allowable maximum derated I_{SPTA} intensity can be increased by a factor of 8 if the ultrasonographer chooses to use the highest available output. A potential 42-fold increase in the maximum allowed I_{SPTA} could be applied in ophthalmic applications, but the MI must be kept below 0.23. The implications for clinical practice are that all users must be aware of the exposure capabilities of high-output devices. There are no automatic safeguards for output of machines that have an ODS set. The user can ensure safety only by keeping the thermal and mechanical indices as low as possible during all examinations.

The Output Display Standard comprises a thermal and a mechanical index that is intended to give a warning of the risk that an extended exposure may produce a significant thermal or non-thermal

bioeffects in human tissue. Temperature rise and cavitation are, in turn, dependent on acoustic factors such as total energy output, the mode, the shape of the ultrasound beam, the position of the focus, the centre frequency, the shape of the waveform, the frame rate and the amount of time during which the beam produces energy. The TI and MI are each designed to take these factors into account, notifying the user about the potential for ultrasound-induced bioeffects [99Abb1].

The TI is a quantity that is related to an estimated temperature rise, and is the ratio of total acoustic power to the acoustic power that would be required to raise tissue temperature by 1°C, under defined assumptions. Assumptions are made about an average value for ultrasonic attenuation for soft tissue. The important acoustic parameters are beamwidth, average power and temporally and spatially averaged intensity. The TI gives a relative estimate of temperature rise at a specific point along the ultrasound beam. This index cannot be assumed to give the actual increase in temperature for all possible conditions, because the assumed conditions for heating in tissue are too complex. A TI of 2 represents a higher temperature rise than a TI of 1, but not necessarily a rise of 2°C. Thus, a theoretical estimate is obtained, based on assumptions about the exposure conditions of tissue properties for any particular examination condition. There are three specific thermal indices: TIS, thermal index for soft tissue; TIB, thermal index for bone; TIC, thermal index for cranial bone. The TIS is used to provide information concerning temperature rise within homogeneous soft tissue, the TIB gives information on temperature rise in bone at or near the focus of the beam and the TIC gives the temperature increase of bone at or near the surface, such as during a cranial examination.

The MI gives a relative indication of the potential for mechanical effects, such as cavitation (the violent collapse of a bubble in tissue). Therefore, the MI is required to be displayed for B-mode imaging. The potential for mechanical effects increases as the peak pressure increases and decreases as the frequency increases for scanning modes. That is probably more significant than thermal effects. According to FDA regulations, the MI can range up to 1.9 for all uses except ophthalmic, for which a maximum MI limit of 0.23 applies. The higher the index value, the higher the probability of a biological effect. The value of learning to implement the ALARA principle allows the use of TI and MI values that are as small as possible, while keeping the quality of the scan as high as possible.

The index that is displayed depends on the choice of transducer, and the mode of examination. For example, for B-mode imaging, the MI will be displayed. For Doppler, M-mode or color flow imaging, the TI will be displayed. Since there are three subcategories of TI (i.e. TIS, TIB and TIC), only one of these is required to be displayed at a time, but the machine must allow the user to retrieve the other two if needed. Only ultrasound systems capable of exceeding an MI or TI of 1 are required to display these indices at values beginning at 0.4 and ranging up to the maximum. Further information is available from the AIUM [94AIU1].

5.4.3.1.3 International recommendations and guidelines

Modern sophisticated ultrasonic equipment is capable of delivering substantial levels of acoustic energy into the body when used at maximum output. The risk of producing bioeffects has been studied by international expert groups during symposia supported by the World Federation for Ultrasound in Medicine and Biology (WFUMB). These symposia have resulted in the publication of internationally accepted conclusions and recommendations. National ultrasound safety committees have published guidelines as well. These recommendations and safety guidelines offer valuable information to help users apply diagnostic ultrasound in a safe and effective manner. However, there is also a modern trend towards self-regulation which has implications for the worldwide use of diagnostic ultrasound. It has resulted in a move away from the relatively simple scheme of FDA-enforced, application-specific limits on acoustic output to a scheme whereby risk of adverse ultrasound exposure is assessed from information provided by the equipment in the form of a real-time display of safety indices. The shift of responsibility for risk assessment from a regulatory authority to the user creates an urgent need for awareness of risk and the development of knowledgeable and responsible attitude to safety issues. To encourage this approach, it is incumbent on authorities, ultrasound societies and expert groups to provide relevant information on biological effects that might result from ultrasonic procedures. It is obvious from the continuous stream of inquiries received by ultrasound societies that effective dissemination of such knowledge requires

sustained strenuous effort on the part of ultrasound safety committees. There is a strong need for continuing education to ensure that appropriate risk/benefit assessments are made by users based on an appropriate knowledge of the probability of biological effects occurring which each type of ultrasound procedure. Concerning the current safety guidelines and international recommendations the actual versions can be downloaded from: <http://www.wfumb.org>, www.aium.org, www.efsumb.org.

The most recent safety recommendations are about the safe use of ultrasound contrast agents, approved by WFUMB Council in October 2005, and a clinical safety statement for diagnostic ultrasound, published by the European Federation of Societies for Ultrasound in Medicine and Biology (EFSUMB) in 2008. The statements are as follows:

WFUMB Safety Committee Recommendations on Contrast Agents (2005, www.wfumb.org)

Clinical Applications and Safety Concerns

- Clinical users should balance the expected clinical benefit from ultrasound contrast agents against the possibility of associated bioeffects.
- Caution should be exercised in the use of microbubble ultrasound in tissues where microvasculature could be dangerous. Some areas of concern include the brain, the fetus and the neonate.
- Clinical users of contrast echocardiography should be alert to the possibility of cardiac rhythm disturbances. Electrocardiograms should be monitored during these procedures.
- Idiosyncratic allergy-like or hypersensitivity reactions are rare but recognized untoward side effects with currently approved agents. Health care professionals are advised to carefully follow the instructions of the package inserts of ultrasound contrast agents. Health care professionals should be alert to the possibility of rare adverse side effects whenever ultrasound contrast agents are administered to patients, and should be prepared to appropriately treat them should any occur.
- The mechanical index (MI) is a useful but imperfect guide for safety and no absolute threshold can be defined. Bioeffects have been observed in small animals in ultrasound contrast agent studies with MI as low as 0.4; the clinical implications are yet to be determined.

Strategies that reduce the likelihood of bioeffects include:

- i. Scanning at lower MI
- ii. Scanning at higher frequencies
- iii. Reducing total acoustic exposure time
- iv. Reducing contrast agent dose
- v. Adjusting the timing of cardiac triggering (end-systole being, in general, the most vulnerable phase for triggering ventricular arrhythmias).

The use of contrast agents in a diagnostic ultrasound study should be avoided 24 hrs prior to lithotripsy procedures.

The clinical significance of microvascular damage and cardiac rhythm effects requires further investigation.

Training

Proper training of new investigators in the clinical use of ultrasound contrast agents is of the utmost importance. Practitioners need to be competent in the administration of contrast agents, familiar with any contra-indications and be able to deal with any possible adverse effect, within the medical and legal framework of their country.

EFSUMB Clinical Safety Statement for Diagnostic Ultrasound (2008, www.efsumb.org)

Diagnostic ultrasound has been widely used in clinical medicine for many years with no proven deleterious effects. However, if used imprudently, diagnostic ultrasound is capable of producing harmful effects. The range of clinical applications is becoming wider, the number of patients undergoing ultrasound examinations is increasing and new techniques with higher acoustic output levels are being introduced. It is therefore essential to maintain vigilance to ensure the continued safe use of ultrasound.

Ultrasound examinations should only be performed by competent personnel who are trained and updated in safety matters. It is important that ultrasound devices are appropriately maintained.

Ultrasound produces heating, pressure changes and mechanical disturbances in tissue. Diagnostic levels of ultrasound can produce temperature rises that are hazardous to sensitive organs and the embryo/fetus. Biological effects of non-thermal origin have been reported in animals but, to date, no such effects have been demonstrated in humans, except when a microbubble contrast agent is present.

The Thermal index (TI) is an on-screen guide to the user of the potential for tissue heating. The Mechanical index (MI) is an on-screen guide of the likelihood and magnitude of nonthermal effects. Users should regularly check both indices while scanning and should adjust the machine controls to keep them as low as reasonably achievable (ALARA principle) without compromising the diagnostic value of the examination. Where low values cannot be achieved, examination times should be kept as short as possible. Guidelines issued by several ultrasound societies are available.

Some modes are more likely than others to produce significant acoustic outputs and, when using these modes, particular care should be taken to regularly check the TI and MI indices. Spectral pulse wave Doppler and Doppler imaging modes (colour flow imaging and power Doppler imaging) in particular can produce more tissue heating and hence higher TI values, as can B-mode techniques involving coded transmissions. Tissue harmonic imaging mode can sometimes involve higher MI values. 3D (three dimensional) imaging does not introduce any additional safety considerations, particularly if there are significant pauses during scanning to study or manipulate the reconstructed images. However, 4D scanning (real-time 3D) involves continuous exposure and users should guard against the temptation to prolong examination times unduly in an effort to improve the recorded image sequence beyond that which is necessary for diagnostic purposes.

Ultrasound exposure during pregnancy

The embryo/fetus in early pregnancy is known to be particularly sensitive. In view of this and the fact that there is very little information currently available regarding possible subtle biological effects of diagnostic levels of ultrasound on the developing human embryo or fetus, care should be taken to limit the exposure time and the Thermal and Mechanical Indices to the minimum commensurate with an acceptable clinical assessment.

Temperature rises are likely to be greatest at bone surfaces and adjacent soft tissues. With increasing mineralisation of fetal bones, the possibility of heating sensitive tissues such as brain and spinal cord increases. Extra vigilance is advised when scanning such critical fetal structures at any stage in pregnancy. Based on scientific evidence of ultrasound-induced biological effects to date, there is no reason to withhold diagnostic scanning during pregnancy, provided it is medically indicated and is used prudently by fully trained operators. This includes routine scanning of pregnant women.

The power levels used for fetal heart rate monitoring (cardiotocography - CTG) are sufficiently low that the use of this modality is not contra-indicated on safety grounds, even when it is to be used for extended periods.

Safety considerations for other sensitive organs

Particular care should be taken to reduce the risk of thermal and non-thermal effects during investigations of the eye and when carrying out neonatal cardiac and cranial investigations.

Ultrasound contrast agents (UCA)

These usually take the form of stable gas filled microbubbles, which can potentially produce cavitation or microstreaming, the risk of which increases with MI value. Data from small animal models suggest that microvascular damage or rupture is possible. Caution should be considered for the use of UCA in tissues where damage to microvasculature could have serious clinical implications, such as in the brain, the eye, and the neonate. As in all diagnostic ultrasound procedures, the MI and TI values should be continually checked and kept as low as possible. It is possible to induce premature ventricular contractions in contrast enhanced echocardiography when using high MI and end-systolic triggering. Users should take appropriate precautions in these circumstances and avoid cardiac examinations in patients with recent acute coronary syndrome or clinically unstable ischaemic heart disease. The use of contrast agents should be avoided 24 hours prior to extra-corporeal shock wave therapy.

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5.5 Summary

Medical Radiological Protection

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Chapter 5 outlines different risks with ionising and non-ionising radiation and outlines some ways of managing risks in clinics. The risks have been shown to be manageable, but continuous work is required in order to achieve reasonable risk levels. Moreover, the health care environment is not static and frequently a new situation has to be handled. Radiation protection in the health care environment is challenging.

The use of ionising and non-ionising radiation for medical purposes has increased during the last decades and more people are exposed to ionising radiation from medical exposures than from any other man-made source. Different medical devices are being developed at a rapid pace. Particularly, the use of computed tomography and magnetic resonance imaging has increased and, today, CT contributes the most to the collective dose in most countries with an advanced health care system. The exposure is unevenly distributed in the population and for specific groups of patients. The radiation dose from diagnostic procedures can be particularly high. Patients outliving radiation therapy are furthermore exposed to significant levels of ionising radiation in non-treated organs. It is important that society, including the health care sector, is aware of this and also assesses the contribution and the risk of late effects for the patient population. Radiation protection in the health care sector is always a hot topic.

Unwanted tissue reactions can be expected for diagnostic, interventional, and therapeutic procedures. This is sometimes unavoidable, but the risk should be known so that affected patients can be treated in a proper manner. Risks should be assessed in order to increase awareness and permit measures to minimise the risk to be taken.

The work at hospitals should be organised in order to minimise occupational exposure to personnel. This can be achieved by planning and evaluation of the work, including personal feed-back to the staff. Even so, hospitals have to be vigilant of the risk of unwanted exposure of staff, especially in radiation therapy.

Some issues that can be highlighted for radiation protection in the health care sector are:

- Patients should not be exposed to unnecessary radiation

The diagnostic and therapeutic procedures applied in the clinic must be carefully chosen. They must be re-evaluated when alternative procedures are available. This is one challenging part of the justification of practices. The appropriateness is complicated to evaluate in the presence of different procedures.

The procedures that are applied have to be optimised in order for the equipment to be used in the most appropriate manner. The optimisation process includes assessment of exposure from non-ionising and radiation to patients and in some cases personnel. In radiotherapy, the dose assessment is an integrated part of the dose-planning process and is applied for the individual patient. In diagnostics in many cases the assessment of exposure is performed for groups of patients. Knowledge of dosimetry is crucial in both diagnostics and therapy.

- Safe transport, storage, handling and use of sources of ionising radiation are needed.

The medical sector uses a large amount of open as well as sealed radioactive sources, many of which could constitute a risk if not properly controlled. This includes a safe handling when out of use.

Some arrangements that must be in place in order to solve radiation protection issues in the clinic are:

- Co-operation between health professionals

The complex issues addressed in this chapter cannot be solved without collaboration between the professionals in the hospital. Expertise in medicine, patient care, and physics are all needed to solve the issues that might arise in the clinic. Thus, collaboration across professional boundaries is essential for a high level of radiation protection.

- Ability to apply radiation protection in the clinic

Knowledge about radiation protection is (easily) obtained with education and training. After such investments there still exists the challenge of applying the knowledge in the clinical work. This would be facilitated by an effective organisation of the clinic and good communication.

Glossary

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The explanations of the terms included in the Glossary are not always identical with definitions as given by national, international or non-governmental organizations such as the ICRP (International Commission on Radiological Protection), the ICRU (International Commission on Radiation Units and Measurements) and the ICNIRP (International Commission on Non-Ionizing Radiation Protection).

Absorbed dose, D

The fundamental dose quantity given by

$$D = \frac{d\bar{\mathcal{E}}}{dm}.$$

where $d\bar{\mathcal{E}}$ is the mean energy imparted to matter of mass dm by ionizing radiation. The SI unit for absorbed dose is joule per kilogram (J kg^{-1}) and its special name is gray (Gy).

Absorbed dose to water

The absorbed dose to water, D_w , is the quotient of $d\bar{\mathcal{E}}$ by dm , where $d\bar{\mathcal{E}}$ is the mean energy imparted to a mass element dm of water, thus

$$D_w = \frac{d\bar{\mathcal{E}}}{dm}.$$

The unit of the absorbed dose is gray (Gy).

Absorbed fraction, $\phi(r_T \leftarrow r_S, E_i, t)$

The fraction of the energy E_i emitted in the source region r_S which is absorbed in the target region r_T at time t .

Absorption coefficient

Absorption coefficient (α) of ultrasound radiation in a medium: The reduction in intensity of an ultrasound beam in the medium, usually expressed in dB cm^{-1} .

Absorption edges

A sharp discontinuity in the absorption spectrum of X-rays by an element that occurs when the energy of the photon corresponds to an orbital energy of the atom.

Absorption, EMF

The dissipation of radiation within a medium, caused by the conversion of radiation.

Absorption, Ionizing Radiation

The termination of the existence of a *photon* through a photoelectric effect or pair production, or of a *particle* by direct collision with a nuclear particle.

Acoustic impedance, Ultrasound

At a given surface, the complex ratio of effective sound pressure averaged over the surface to the effective flux (volume velocity or particle velocity multiplied by the surface area) through it. The unit is the Nsm^{-1} (newton-second/meter). Acoustic impedance, being a complex quantity, can have real and imaginary components analogous to those in an electrical impedance. In applying this analogy, the real part of the acoustic impedance is termed acoustic resistance, and the imaginary part is termed acoustic reactance.

Acoustic pressure

In ultrasound, the instantaneous value of the total pressure minus the ambient pressure. Unit: pascal (Pa).

Action potential, Neurophysiology

An action potential is a brief change in membrane potential (depolarisation) caused by the rapid opening and closing of ion channels thus allowing passage of specific ions. APs travel along axons and transfer information over distance.

Added filtration

Removable or irremovable filter positioned in the radiation beam of an X-ray tube.

Adverse health effect

A biological effect which has a detrimental effect on mental, physical and/or general well being of an individual due to exposure to an electric field, a magnetic field, an electromagnetic field, a contact current, or to ionizing radiations, either in the short term or in the long term.

Afterloading technique

Manually or remotely controlled transfer of one or more sealed radioactive source(s) between a storage container and a pre-positioned source applicator for brachytherapy.

Air kerma, K_a

Air kerma, K_a , is the quotient of ΔE_{tr} by Δm , where ΔE_{tr} is the sum of the initial kinetic energies of all the charged particles liberated by uncharged particles in a mass of air of Δm_a , thus

$$K_a = \frac{\Delta E_{tr}}{\Delta m_a}.$$

The unit of kerma is gray (Gy).

Air kerma area product (KAP)

Product of the area of a cross-section of a radiation beam and the average value of the *air kerma* over that cross-section. This quantity is available clinically either by direct measurement with a KAP meter or by calculator and display on a KAP indicator.

Air kerma strength, S_K

Air kerma strength, S_K , is the *air kerma* rate, $\dot{K}_\delta(d)$, *in vacuo* and due to photons of energy greater than δ , at distance d , multiplied by the square of this distance, d^2 ,

$$S_K = \dot{K}_\delta(d) \cdot d^2.$$

AMAD

Activity Median Aerodynamic Diameter (μm). This means in inhalation of radioactive particles that half of the activity is related to particles with a higher aerodynamic diameter and the other half to particles with a lower aerodynamic diameter.

Ambient dose equivalent

$H^*(d)$, an operational quantity at a point of interest in the real radiation field, is the dose equivalent that would be produced by the corresponding aligned and expanded radiation field, in the ICRU sphere at a depth d , on the radius vector opposing the direction of radiation incidence. For strongly penetrating radiation it is $d = 10$ mm and $H^*(d)$ is written $H^*(10)$. The unit of ambient dose equivalent is sievert (Sv).

Ampere

Named after André-Marie Ampère, 1775-1836. SI-Unit of the electric current, 1 ampere (A) equals the flow of one coulomb of charge each second.

Annihilation radiation

Photon radiation produced by the recombination of an electron and a positron.

Anode angle

The angle between the directions of the electron beam in an X-ray tube and the normal to the anode surface

Anthropomorphic phantom

Phantom with structures and contrast details designed to mimic (parts of) the human body.

Apoptosis

Cell biology. An active process requiring metabolic activity by the dying cell, characterised by enzymatic degradation of the cell DNA and proteins without inflammatory responses. Apoptosis is induced by several factors not yet understood totally. It is interpreted as protection of the organism by removing damaged, useless or overaged cells.

Athermal

Synonym: non-thermal. Not pertaining to heat or not caused by heat.

Atomic cross section

Describes the probability of an interaction between matter/absorber and incident radiation.

Atomic mass unit

$u = 1.660\ 538\ 782 \times 10^{-24}$ g.

Attenuation coefficient

In radiography or ultrasound, the difference between the energy that enters a body part and the energy that is not detected. The difference is caused by the absorption and scattering of energy within the body tissues. It is usually denoted as α and measured in dB/cm/MHz.

Attenuation-equivalent filtration

Thickness of a layer of reference material which, if substituted for the material under consideration in a beam of specified radiation quality, gives the same degree of attenuation.

Auger electron

Possible result of de-excitation of the atomic shell following an ionization process is called after one of its discoverers, Pierre Victor Auger. See also *Characteristic X-rays*.

Backscatter factor, B

The ratio of the entrance surface *air kerma* to the incident *air kerma*.

Basic restrictions, EMF

Restrictions on exposure to time-varying electric, magnetic, and electromagnetic fields that are based directly on established health effects are termed “basic restrictions”. Depending upon the frequency of the field, the physical quantities used to specify these restrictions are current density (J), specific energy absorption rate (SAR), and power density (S). Only power density in air, outside the body, can be readily measured in exposed individuals.

Beam quality correction factor, k_Q

Correction factor, which accounts for the change in response in terms of absorbed dose to water between the radiation quality of interest, Q , and the reference radiation quality for which the dosimeter's calibration factor applies (usually ^{60}Co). The beam quality correction factor k_Q is a function of the beam specifier Q and is chamber dependent. For ^{60}Co beams $k_Q=1.000$.

Beam specifier

A parameter by means of which the radiation quality is characterized. In the case of high energy bremsstrahlung, electron radiation and hadron radiation the tissue-phantom ratio, R_{50} and the residual range, respectively, are frequently used. R_{50} denotes the depth inside in a water phantom in which the dose rate has half of its value at the dose maximum.

Biological effect

A measurable change in a biological system in response (for example) to an electromagnetic field or ionizing radiation.

Biological treatment planning

Extends the physical treatment planning in radiation oncology to the different biological impact for different radiation qualities at the same absorbed dose.

Biologically effective dose (BED)

The uniform absorbed dose delivered at a reference dose rate that would lead to the same effects as the actual dose. See also *Equivalent uniform dose (EUD)*.

Biophysical mechanism

Physical and/or chemical interactions of electric fields, magnetic fields, electromagnetic fields, or ionizing radiation with biological systems.

Blood-brain barrier

A physiological “barrier” comprising endothelial and epithelial cells that regulates the composition of cerebrospinal fluid of the central nervous system.

Brachytherapy

Internal radiotherapy, a form of radiotherapy where a radiation source is placed inside or next to the area requiring treatment. Brachytherapy is commonly used as an effective treatment for cervical, prostate, breast and skin cancer and can also be used to treat tumours in many other body sites.

Bragg peak

A pronounced peak on the depth dose curve for beams of protons or heavier ions immediately before the particles come to rest, called after William Henry Bragg who described it in 1903.

Bragg-Gray principle

Idealized condition of a field of photon or electron (charged particles) radiation in and around a gas filled cavity embedded in a liquid or solid material assuming that the field of electrons (charged particles) is essentially not modified by the presence of the cavity.

Bremsstrahlung

Photon radiation produced through the deceleration of electrons in the field of atomic nuclei. Bremsstrahlung exhibits a broad smooth spectral distribution.

Broad beam geometry

Form of irradiation employing a laterally extended beam. A broad beam geometry is usually accompanied by the occurrence of scattered radiation.

Bucky tray

A sliding metal tray under the radiographic table which holds the grid and the film or screen cassette.

Build-up

Phenomenon of the increase with depth of the absorbed dose rate due to the release of secondary charged particles and to scattered radiation in matter beyond the entrance surface.

Calibration

Determination of the calibration factor, N , under reference conditions. The indication, M , of the dosimeter corrected to standard air density is related to the dose, D , at the point of test by

$$N = \frac{D}{M}.$$

Here, dose is a generic term denoting either air kerma or absorbed dose to water. It has been determined with a primary or secondary standard.

Calibration factor, N

For a detector assembly with an associated measuring assembly, the calibration factor converts the indication, corrected to stated reference conditions, to the conventional true value of the dosimetric quantity at the reference point of the detector. When the display unit indicates the dosimetric quantity together with its appropriate unit, the calibration factor is dimensionless. For a detector calibrated on its own without a specified measuring assembly, the calibration factor converts the output (usually current or

charge), corrected to reference conditions, to the dosimetric quantity at the reference point of the detector. In this case the calibration factor has a dimension, e.g. gray/coulomb, a situation in which it is also referred to as calibration coefficient.

Calibration quality

Radiation quality employed for the calibration of dosimeters. For radiotherapy it is usually the radiation of ^{60}Co , in diagnostic radiology a X-ray quality is used.

Cancer

General term for an uncontrolled and abnormal malignant growth of cells (neoplasm, tumor).

Carcinogen

An agent that can induce cancer.

Carcinoma

A malignant new growth made up of epithelial cells, found in skin or the lining of body organs, e.g., in breast, lung, stomach or bowel. Carcinomas tend to infiltrate the surrounding tissues and spread to distant organs (metastases). Carcinomas make up 80% of all malignant tumours.

Causal relationship

A causal relationship occurs between two agents when one causes the other. For example, researchers are studying whether there is a causal relationship between EMF or ionizing radiation and cancer, meaning that they are studying to see if *EMF* or ionizing radiation causes, or affects the progress of, cancer.

Cavitation

Cavitation is defined as the phenomenon of formation of vapour bubbles of a flowing liquid in a region where the pressure of the liquid falls below its vapour pressure. Cavitation is usually divided into two classes of behaviour: inertial (or transient) cavitation and non-inertial cavitation. Inertial cavitation is the process where a void or bubble in a liquid rapidly collapses, producing a shock wave. Such cavitation often occurs in pumps, propellers, and in the vascular tissues of plants. Non-inertial cavitation is the process in which a bubble in a fluid is forced to oscillate in size or shape due to some form of energy input, such as an acoustic field. Such cavitation is often employed in ultrasonic cleaning baths and can also be observed in pumps, propellers, etc.

Cell signalling (pathways)

A sequence of intracellular changes linking a “signalling event”, such as activation of membrane-bound ion channels or ligand-receptors, and a “response”, such as a change in gene expression, for example, leading to increased proliferation.

Central nervous system (CNS)

Usually taken to mean the cells, such as neurons or glial cells, of the brain and spinal cord. It also includes the retina, which is formed as an outgrowth of the forebrain.

Characteristic X-rays

Monoenergetic photons as a result of de-excitation of the atomic shell, see also *Auger electron*.

Chromosome

In a prokaryotic cell or in the nucleus of an eukaryotic cell, a chromosome is a structure that carries the genetic information. A chromosome consists of one long DNA molecule with the genes in linear sequence, and associated RNA and proteins. At beginning of nuclear division they are easily visible under a microscope after staining.

Chronic effect

Consequence which develops slowly and has a long-lasting course (often but not always irreversible).

Chronic exposure

Exposure lasting for very long periods, e.g. significant periods of the working life.

Cognition

Information processing by the brain, including processes such as attention, perception, learning, reasoning, comprehending and memory.

Coherent scattering

Photon – electron interaction in which the photon undergoes a change in direction without a net exchange of energy.

Collapsed Cone

A method for photon beam dose calculations that overcomes some of the limitations of the simpler Pencil Beam approach by fully accounting for inhomogeneities present in the irradiated volume.

Collision mass-stopping power

Linear collision stopping power divided by the density of the material.

Collision Stopping Power

Description of the interaction of charged particles with the atomic shell, see also *Radiative mass-stopping power*. The term is also referred to as *linear collision stopping power*.

Committed equivalent dose, $H(r_T, T_D)$

The integration of the equivalent dose rate to the target tissue r_T up to time T_D .

Compact chamber

Ionization chamber in which the outer electrode takes the form of a rigid cylindrical wall closed at one end and mounted at the other on the supporting stem. This type of chamber is also referred to as thimble chamber.

Compartment model

A mathematical model with compartments and the description of transfer processes between the compartments. They are used for the description of biokinetic models when the compartments represent source regions and the transfer of activity to other source regions is described by transfer rates.

Compton electron

Electron liberated from the atomic shell by a Compton scattering process.

Compton scattering, Compton effect

Photon – electron interaction in which part of the photon's energy is transferred to the electron. The process is governed by energy and momentum conservation.

Computed tomography (CT)

An imaging method employing tomography created by computer processing. Digital geometry processing is used to generate a three-dimensional image of the inside of an object from a large series of two-dimensional X-ray images taken around a single axis of rotation. CT produces a volume of data which can be manipulated in order to demonstrate various bodily structures based on their ability to block the X-ray beam. Historically the images generated were in the axial or transverse plane, but modern scanners allow this volume of data to be reformatted in various planes or even as volumetric (3D) representations of structures.

Computed Tomography Dose Index (CTDI)

The integral of the air kerma $K(z)$ along an axis parallel to the rotational axis of the scanner for one scanner's rotation, divided by the product of the number of tomographic sections and the nominal tomographic section thickness

$$\text{CTDI} = \frac{1}{N \cdot h} \int_{-50\text{mm}}^{+50\text{mm}} K(z) dz .$$

As the pencil shaped CT chambers have usually a measuring volume of length of 100 mm the integration is carried out over this length. The CTDI is either measured free in air on the axis of a scanner or inside the so-called body or head phantom. These are polymethyl-metacrylate cylinders with diameters of 16 cm

and 32 cm equipped with a hole along the axis and holes 1 cm below the periphery parallel to the axis to position the pencil shaped CT chambers. The CTDI is usually given in units of mGy.

Conductivity, electrical

The scalar or vector quantity which, when multiplied by the electric field strength, yields the conduction current density. It is the reciprocal of resistivity. Expressed in siemens per meter (S m^{-1}).

Conjugate counting

A technique for quantitatively measuring the amount of radioactivity by positioning two detectors registering the radiation from opposite directions. By taking the geometric mean a quantity is obtained that is independent of the position of the source.

Constant potential

High voltage supplied to an X-ray tube with a ripple not exceeding a specific value, frequently an upper limit of 10% is used.

Contact current

Current flowing between an energized, isolated, conductive (metal) object and ground through an electrical circuit representing the equivalent impedance of the human body.

Continuous wave

A wave whose successive oscillations are identical under steady-state conditions.

Controlled operating mode

Because of the uncertainty over identified deleterious effects during MRI procedures, it is recommended that exposure limits for the patients be divided into three tiers. The controlled operating mode is used for specific MR examinations outside the normal operating range where discomfort and/or adverse effects for some patients may occur. A clinical decision must be taken to balance such effects against foreseen benefits; exposure must be carried out under medical supervision.

Conversion coefficient

Usually a dimensional parameter which allows to convert a given physical quantity to another one. An example is the parameter mass-energy transfer coefficient μ_{tr}/ρ relating the (photon) energy fluence Ψ to the kerma $K = (\mu_{\text{tr}}/\rho)\Psi$.

Conversion of pre-neoplastic cells

Pre-neoplastic cells convert to a form, in which they are committed to become fully malignant.

Cortical bone

In the ICRP models cortical bone are those bones with a surface/volume ratio less than 60 cm^{-1} . It is also called compact bone.

Cross section, σ

The cross section of a target entity, for a particular interaction produced by incident particles is the quotient of P by Φ , where P is the probability of that interaction for a single target entity when subjected to the particle fluence Φ . It is $\sigma = P/\Phi$, unit: m^2 . A special unit often used for the cross section is the barn (b) with $1 \text{ b} = 10^{-28} \text{ m}^{-2}$.

Cross-calibration

Calibration of a dosimeter by means of a calibrated reference dosimeter in the users beam. Electron dosimetry with parallel-plate chambers frequently resorts to a cross-calibration.

CT pitch factor

In helical scanning the ratio of the patient support travel Δd along the z direction per rotation of the X-ray source divided by the product of the nominal tomographic section thickness T and the number of tomographic sections N :

$$\text{CT pitch factor} = \frac{\Delta d}{N \cdot T}.$$

Current

The movement of electrons in a conductor.

Current density

A vector of which the integral over a given surface is equal to the current flowing through the surface; the mean density in a linear conductor is equal to the current divided by the cross-sectional area of the conductor. Unit: ampere per square meter (A m^{-2}).

Cyclotron

A type of particle accelerator used in radiation oncology to produce heavy charged particle beams, see also *Linear accelerator* and *Synchrotron*.

DC

Abbreviation for “Direct Current”, but also used to indicate constancy of fields, see *Static field*.

Decibel (db)

A measure of the increase or decrease in power, P , at two points expressed in logarithmic form. Gain: $10 \log_{10}(P_2/P_1)$.

Delta rays

Secondary electrons produced in an interaction with an energy sufficiently high to allow ionization and excitation processes. By definition, the secondary electron in electron-electron interactions may have up to half of the kinetic energy of the primary electron.

Densely ionizing radiation

Categorizes a radiation quality in regard to its energy deposition per unit length. A rather arbitrary limit at 3-5 keV/ μm separates sparsely ionizing (energy transfer below 3-5 keV/ μm) from densely ionizing radiation.

Density of air correction

Multiplication of the reading of a vented ionization chamber with the ratio of the reference air density (stated in the calibration certificate, usually 1013.25 hPa, 20°C) to that actually prevailing at the time of measurement.

Depth dose curve

Graphical presentation of the variation of the dose along the beam axis as a function of the depth inside a phantom.

Deterministic biological effects

Differentiated cells have developed specific morphology and function and are predestined to die. The rate of death of differentiated cells, in healthy state of an individual is balanced by proliferation from stem cells. If enough stem cells in a tissue are killed, e.g. by irradiation from ionizing radiation, or prevented from undergoing cell division there will be loss of tissue function termed in radiological protection by the ICRP as deterministic.

Deterministic effects are characterised by a threshold dose and an increase in the severity of the reaction as the dose is increased further. In some cases, deterministic effects are modifiable by post-irradiation procedures including biological response modifiers.

Detriment

The total harm to health experienced by an exposed group and its descendants as a result of the group's exposure to a radiation source. Detriment is a multidimensional concept. Its principal components are the stochastic quantities: probability of attributable fatal cancer, weighted probability of attributable non-fatal cancer, weighted probability of severe heritable effects, and length of life lost if the harm occurs.

Diagnostic Reference Level (DRL)

Used in medical imaging with ionizing radiation to indicate whether, in routine conditions, the patient dose or administered activity (amount of *radioactive* material) from a specified procedure is unusually high or low for that procedure.

Dielectric constant

See *Permittivity*.

Dipole

A centre-fed open antenna excited in such a way that the standing wave of current is symmetrical about the mid point of the antenna.

Direct effect

A biological effect resulting from direct interaction of EMF or ionizing radiation with biological structures.

Direct measurement

Dose measurements in diagnostic radiology carried out during the examination of the patient or during the exposure of a phantom.

Displacement effect

Deviation of a real detector from ideal Bragg-Gray conditions caused by the finite extension of the detector resulting in a modification of the undisturbed radiation field.

DNA (deoxyribonucleic acid)

This polymeric molecule (consisting of deoxyribonucleotide building blocks in a double-stranded, double-helical form) is the genetic material of most organisms.

Dose and dose rate effectiveness factor (DDREF)

Factor to project cancer risk determined at high doses and dose rates of ionizing radiations to the risks that could apply at low doses and low dose rates.

Dose coefficient (mGy/MBq or mSv/MBq)

The absorbed organ dose or the effective dose per administered unit activity.

Dose constraint

A prospective and source-related restriction on the individual dose from a source, which provides a basic level of protection for the most highly exposed individuals from a source, and serves as an upper bound on the dose in optimisation of protection for that source. For occupational exposures, the dose constraint is a value of individual dose used to limit the range of options considered in the process of optimisation. For public exposure, the dose constraint is an upper bound on the annual doses that members of the public should receive from the planned operation of any controlled source. Dose constraints do not apply to medical exposures of patients; *diagnostic reference levels* are used in the management of patient doses in medical examinations.

Dose limit

The value of the *effective dose* or the *equivalent dose* to individuals from planned exposure situations that is not to be exceeded.

Dose reference level

See *Diagnostic Reference Level (DRL)*.

Dose-length product

(a) Generic term to denote the line integral of the air kerma along the z-axis of a CT scanner.
(b) Product of the weighted CTDI normalised to tube loading, ${}_{\text{n}}\text{CTDI}_{\text{w}}$, tube loading per single slice (mAs), W , total number of slices imaged, n , and nominal slice thickness or detector group in multi-slice scanning, T :

$$\text{DLP} = {}_{\text{n}}\text{CTDI}_{\text{w}} \cdot W \cdot n \cdot T .$$

Dosimetry, EMF

Measurement, or determination by calculation, of internal electric field strength or induced current density, of the specific energy absorption, or specific energy absorption rate distribution, in humans or animals exposed to electromagnetic fields.

Effective dose (Ionizing radiation), E

According to the MIRD technology (which is basically the same as that of ICRP) the tissue-weighted sum of the equivalent doses in all specified tissues and organs of the body is given by the expression:

$$E = \sum_T w_T \cdot H(r_T), \text{ with } H(r_T) = \sum_R w_R \cdot D_R(r_T) \text{ and } \sum_T w_T = 1,$$

where $H(r_T)$ is the equivalent dose absorbed in a target region (tissue or organ) r_T , and w_T is the tissue weighting factor. In the expression for $H(r_T)$, $D_R(r_T)$ denotes the mean dose absorbed in the target region due to radiation of type R and radiation weighting factor w_R . The unit for the effective dose is the same as for absorbed dose, J kg^{-1} , and its special name is sievert (Sv).

Effective irradiance (Non-ionizing radiation), E_{eff}

When considering health effects of optical radiation, an effective exposure rate (i.e., effective irradiance) E_{eff} is calculated by spectral weighting as follows: the spectral irradiance E_λ at the surface of the exposed biological tissue is mathematically weighted against the action spectrum of the biological response $A(\lambda)$ across the relevant spectrum. Unit: watt per square meter (W m^{-2}).

Effective point of measurement

In cavity theory that point in a solid or liquid phantom of material m in which the absorbed dose to material m is equal to the average absorbed dose to the gas, g , in the cavity of a detector multiplied by the ratio of the averaged mass-stopping powers of material m to cavity gas g . The effective point of measurement is usually not at the centre of the cavity but a certain distance 'up-stream' towards the radiation source. When the distance between the effective point of measurement and the cavity centre is known, it is convenient to define the effective point of measurement in the coordinate system of the detector.

Effective radiant exposure, H_{eff}

The effective radiant exposure H_{eff} (or effective exposure dose) is the product of the exposure duration t , in seconds, and the effective irradiance E_{eff} (spectrally weighted optical radiation). Unit: joule per square meter (J m^{-2}).

Elastic scattering

Interaction process in which the kinetic energy is conserved.

Electric charge, Q

The quantity of electricity Q is the integral of electric current over time. Unit: coulomb (Q).

Electric current

Electric current is the flow of electricity. A voltage will always try to drive a current. The size current that is driven depends on the resistance of the circuit.

Electric field

A vector field E measured in volts per meter (V m^{-1}).

Electric field strength, E

Force exerted by an electric field on an electric point charge, divided by the electric charge. Expressed in newtons per coulomb or volts per meter ($\text{N C}^{-1} = \text{V m}^{-1}$).

Electric flux density, D

The electric flux density (or dielectric displacement), a vector quantity, the divergence of which is equal to the volume density of charge. The electric flux density is related to the electric field strength by the properties of the medium in a given location r as $D(r) = \epsilon E(r)$, where ϵ is the complex permittivity. Unit: coulomb per square meter (C m^{-2}).

Electromagnetic interference

Degradation of the performance of a device, a piece of equipment, or a system caused by an electromagnetic disturbance.

Electron volt

The kinetic energy gained by an electron by running through a potential difference of one volt, symbol eV: $1 \text{ eV} = 1.602\,176\,487 \times 10^{-19} \text{ J}$

ELF

Extremely low frequency; frequency in the range between 1-300 Hz.

Embryo

The stage of prenatal development between the fertilised ovum and the completion of major organ development. In humans, this occurs in the first trimester.

EMF

Electric, magnetic, and electromagnetic fields.

Energy fluence rate

Derivative of the energy fluence with respect to time.

Energy fluence, Ψ

Quotient of dR by da , where dR is the radiant energy incident on a sphere of cross-sectional area da . $\Psi = dR / da$, Unit: J m^{-2} .

Entrance surface air kerma

The air kerma at a point in a plane corresponding to the entrance surface of a specified object, e.g. a patient's breast or a standard phantom. The radiation incident on the object and the backscattered radiation are included.

Entrance surface dose

Absorbed dose to tissue, including the contribution from backscatter, assessed at a point on the entrance surface of a specified object.

Epidemiological study

The study of the distribution of disease in population and of the factors that influence this distribution.

Equivalent dose, H_T

According to the MIRD technology (which is basically the same as that of ICRP) the dose in a tissue or organ T is given by the expression:

$$H(r_T) = \sum_R w_R \cdot D_R(r_T),$$

where $D_R(r_T)$ is the mean absorbed dose from radiation R in a target region (tissue or organ) r_T , and w_R is the radiation weighting factor. Since w_R is dimensionless, the unit for the equivalent dose is the same as for absorbed dose, J kg^{-1} , and its special name is sievert (Sv).

Equivalent uniform dose (EUD)

The uniform absorbed dose that would lead to the same effect as a nonuniform dose distribution. See also *Biologically effective dose (BED)*.

Established mechanism

A bioelectric mechanism having the following characteristics:

- (a) can be used to predict a biological effect in humans;
- (b) an explicit model can be made using equations or parametric relationships;
- (c) has been verified in humans, or animal data can be confidently extrapolated to humans;
- (d) is supported by strong evidence; and
- (e) is widely accepted among experts in the scientific community.

Excretion

The removal of a substance from the body, mainly via the urinary bladder to urine or via the intestines to faeces.

Experimental operating mode, MRI

Because of the uncertainty over identified deleterious effects during MRI procedures, it is recommended

that exposure limits for the patients be divided into three tiers. The experimental operating mode is used for experimental MR procedures, at levels outside the controlled operating range, for which special ethical approval is required in view of the potential risks.

Exposure, EMF

The subjection of a person to electric, magnetic, or electromagnetic fields or to contact currents other than those originating from physiological processes in the body and other natural phenomena.

Exposure, Ionizing Radiation

(a) Physical quantity for in-air dose measurements. It has been superseded by the air kerma. Air kerma K and exposure J are related to each other by $J = K(e/W)$, where the term in parenthesis is the elemental charge divided by the W -value.

(b) See *Patient dose (exposure)*.

Exposure metric, EMF

A single number that summarizes an electric and/or magnetic field exposure over a period of time. An exposure metric is usually determined by a combination of the instrument's signal processing and the data analysis performed after the measurement.

Exposure parameters

The settings of X-ray tube voltage (kV), tube current (mA) and exposure time (s).

External dosimetry

Dose assessment in the case of irradiation of the body from external radiation sources, for example in the case of X-ray examinations.

External radiation therapy

Form of radiation therapy in which the radiation source is outside the body. This form of treatment is also referred to as teletherapy.

Extremities

Limbs of the body.

Far field

The region where the distance from a radiating antenna exceeds the wavelength of the radiated EMF; in the far-field, field components (E and H) and the direction of propagation are mutually perpendicular, and the shape of the field pattern is independent of the distance from the source at which it is taken.

FDG

Fluorodeoxyglucose, labelled with ^{18}F is the most frequently used PET radiopharmaceutical.

Fetus (foetus)

The stage of prenatal development between the embryo and birth.

Fibrillation (ventricular)

The loss of organised ventricular contractions of the heart.

Filtration

Modification of characteristics of ionizing radiation on passing through matter. Filtration may be (a) a preferential absorption of certain components of polyenergetic X-rays or gamma radiation accompanying its attenuation, or (b) a modification of the distribution of radiation intensity over the cross-section of a radiation beam.

Fluence ϕ

Quotient of dN by da , where dN is the number of particles incident on a *sphere* of cross-sectional area da .

$$\phi = dN / da, \text{ unit: m}^{-2}.$$

Fluence rate

Derivative of the fluence with respect to time.

Fluoroscopically guided interventional procedures

Procedures comprising guided therapeutic and diagnostic interventions, by percutaneous or other access, usually performed under local anaesthesia and/or sedation, with fluoroscopic imaging used to localise the lesion/treatment site, monitor the procedure, and control and document the therapy.

Flux

\dot{N} is the quotient of dN by dt , where dN is the increment of the particle number in the time interval dt .

$$\dot{N} = \frac{dN}{dt}, \text{ unit: s}^{-1}$$

Fractional absorption

The fraction of material which is absorbed to the blood from the alimentary tract (mostly from the small intestine). It is given as the f_1 value in the old gastro-intestinal tract model and as the f_A value in the Human Alimentary Tract Model.

Fractional uptake

The fraction of material in blood which is taken up by a specified organ.

Free air chamber

Ionization chamber used for the realization of the unit of the air kerma for photon radiation with energies up to about 500 keV. An essential feature of a free air chamber is the establishment of secondary electron equilibrium in the measurement volume.

Free radicals

Highly reactive chemical species (part of a molecule) with an unpaired electron.

Frequency

The number of cycles completed by electromagnetic waves in 1 s; usually expressed in hertz (Hz).

Gaf-chromic film

Film whose optical density is increased by being exposed to ionizing radiation. It consists of a sensitive layer sandwiched between two coatings which polymerizes when exposed to radiation. The film is also referred to as radiochromic film.

Gamma camera

Standard equipment for nuclear medicine imaging. Gamma radiation is registered by a detector in the form of a large (typically 40 cm × 50 cm, 1 cm thick) NaI(Tl) scintillation detector. In front of the detector is a collimator, that allows only radiation perpendicular to the detector to pass. For the location of the scintillations in the detector a large number of PM-tubes is used.

Gastrointestinal syndrome

Radiation induced syndrome of the gastrointestinal tract after acute irradiation of the small intestine with absorbed doses of a syndrome threshold of about 3 Gy. Major symptoms are vomiting, diarrhoea with bleeding, and disturbed electrolyte and fluid balance.

Gene expression

The production of a functional protein or an RNA molecule from genetic information (genes) encoded by DNA.

General recombination

Recombination of charge carriers of opposite sign generated by ionizing radiation on their way from the point of production to their respective collecting electrodes in the electrical field of an ionization chamber.

Geomagnetic field

Magnetic field originated from the Earth (including the atmosphere). Predominantly a static magnetic field, but includes some oscillating components and transients.

Gradient magnetic field

A magnetic field that is not spatially uniform. The rate at which the magnetic field changes as a function of location, which is measured in units of tesla per meter ($T\ m^{-1}$).

Guidance level, diagnostic reference level

A level of a specified quantity above which appropriate action should be considered. In some circumstances, action may need to be considered when the specified quantity is substantially below the guidance or reference level.

Guideline

A recommended limit for a substance or agent intended to protect human health or the environment.

Haematology

The study of blood; its formation, normal composition, function and pathology.

Haematopoietic syndrome

Radiation induced syndrome of the haematopoietic system after acute irradiation of the red bone marrow with absorbed doses of a syndrome threshold dose of about 2 Gy. Major symptoms are fever, internal bleeding, depletion of bone marrow leading to lower blood counts.

Half-life (effective)

The time period during which the activity in specified organs or tissues is decreased to 50%. The effective half-life takes into account the biological half-time and the physical half-life.

Half-time (biological)

The time period during which the material in specified organs or tissues is decreased to 50%. For radioactive material the decay is not considered.

Half-value layer (HVL)

Thickness of a specified material which attenuates under narrow beam conditions X-rays or gamma radiation to an extent such that the air kerma rate is reduced to one half of the value that is measured without the material. This quantity is also referred to as first HVL. The second HVL is the thickness of additional material to attenuate the air kerma from one half to one quarter of the value in the unattenuated beam.

Hardening-equivalent filtration

Thickness of a layer of reference material which, if substituted for the material under consideration in a beam of specified radiation quality, gives the same degree of beam hardening

Harmonic

A frequency which is a multiple of the frequency under consideration.

HATM, human alimentary tract model

A mathematical model describing the movement of material through the alimentary tract from the oral cavity via the oesophagus, the stomach, and the intestines to faecal excretion as well as the absorption of material from the tract to blood.

Health

A state of complete physical, mental and social well-being and not merely the absence of disease or infirmity.

Health hazard

A biological effect that is detrimental to health or well-being.

Heart rate

The measurement of the number of heartbeats per minute.

Hertz (Hz)

The unit for expressing frequency, f . One hertz equals one cycle per second. 1 kHz = 1000 Hz, 1 MHz = 1000 kHz, 1 GHz = 1000 MHz.

Homogeneity coefficient

Ratio of the first to the second half-value layer. For poly-energetic radiation its value is always less than one.

HRTM, human respiratory tract model

A mathematical model describing the deposition of inhaled material in the respiratory tract and its removal by extrinsic means to the environment, mechanical clearance to the alimentary tract and lymph nodes, and by absorption to blood.

Hyperthyroidism

Term for overactive tissue within the thyroid gland causing an overproduction of thyroid hormones.

Image intensifier

Electronic device used in fluoroscopy for the conversion of the radiation relief into a visible image.

Immune system

The body's primary defence against abnormal growth of cells (i.e. tumours) and infectious agents such as bacteria, viruses, and parasites.

Impedance (of free space)

Ratio of electric to magnetic field strength of an electromagnetic wave. In free space the value is 377Ω .

In situ

Within biological tissue.

In vitro

Experimental studies of cells or tissues, usually in a sustaining oxygenated, fluid medium. Literally means "in glass", isolated from the living organism and artificially maintained, as in a test tube or culture dish.

In vivo

Occurring within the whole living body. "In life"; experimental studies of processes in living organisms.

Incident air kerma

The air kerma at a point in a plane corresponding to the entrance surface of a specific object, e.g., a patient's breast or a standard phantom. Only the radiation incident on the object and not the backscattered radiation is included.

Incomplete charge collection

Loss of charge carriers of opposite sign produced by ionizing radiation through the processes of recombination. Usually a distinction is made between initial and general recombination, where the first term refers to those processes occurring within the track of a (primary) single charged particle. The second term accounts for all other effects of recombination.

Indirect effect, EMF

An effect on the human body resulting from the interaction between an electromagnetic field and another object, such as a vehicle or other mechanical structure, with which the body comes into contact, e.g., shock or burn.

Indirect measurement

Dose measurements in diagnostic radiology based on the setting of X-ray tube voltage and current and on the geometrical conditions of the exposure.

Induction

An electric or magnetic field in a conducting medium caused by the action of a time-varying external (environmental) electric or magnetic field.

Inelastic scattering

Interaction process of ionizing radiation in which a transformation from one form of energy to another one occurs, e.g., in the case of pair production.

Influence quantity

Any external quantity that itself is not the subject of the measurement, but which may affect the performance of a measuring instrument. Influence quantities may be external in nature, e.g., ambient temperature, humidity, etc., or internal like, e.g., battery voltage.

Infrared radiation

Electromagnetic radiation capable of producing the sensation of heat and found between visible radiation and radiofrequency radiation in the electromagnetic spectrum (wavelengths 700 nm to 1000 μm).

Inherent filtration

The filtration provided by permanent materials through which the radiation beam must pass before emerging from the X-ray tube.

Initial recombination

Recombination of charge carriers of opposite sign generated within the track of a single (energetic) charged particle within the ionized medium.

Instantaneous

Adjective used to describe particular parameters that must be measured or evaluated over a very short time interval (typically 100 microseconds or less).

Intensity Modulated Radiation Therapy (IMRT)

Form of external radiation therapy in which for each direction of the incident beam the particle fluence over the cross-sectional area of the beam is deliberately varied with the objective of obtaining the highest possible degree of conformity between the target volume and the dose profile.

Interaction

Reaction based on the mutual forces between particles and atoms or atomic nuclei or between colliding particles by means of which the energy and/or direction of the incoming particle is modified.

Internal dosimetry

Dose assessment after incorporation of radionuclides, for example in nuclear medicine after administration of radiopharmaceuticals.

Ionization chamber

Detector consisting of a chamber filled with a suitable medium, usually gaseous, in which an electric field, insufficient to induce charge multiplication, is provided for the collection at the electrodes of charges associated with ions and the electrons produced in the sensitive volume of the detector by ionizing radiation. The ionization chamber includes the sensitive volume, the collecting and polarizing electrodes, the guard electrode (if any), the chamber wall, the parts of the insulator adjacent to the sensitive volume and any necessary build-up caps to ensure electron equilibrium.

Irradiance

Irradiance describes the flux, radiative power density, and incidence on a surface. Units: W m⁻² or W cm⁻². The surface must be specified for the irradiance to have meaning.

Justification

The process of determining whether either

- (1) a planned activity involving radiation is, overall, beneficial, i.e., whether the benefits to individuals and to society from introducing or continuing the activity outweigh the harm (including radiation detriment) resulting from the activity; or
- (2) a proposed remedial action in an emergency or existing exposure situation is likely, overall, to be beneficial. In the context of medical exposures of patients, justification is always of the first kind.

Kerma

The kerma, K , is the quotient of ΔE_{tr} by Δm , where ΔE_{tr} is the sum of the initial kinetic energies of all the charged particles liberated by uncharged particles in a mass of Δm , thus

$$K = \frac{\Delta E_{\text{tr}}}{\Delta m}.$$

The special name for the unit of kerma is gray (Gy).

Note 1: the kerma has the special name *air kerma* when the mass element Δm consists of air.

Note 2: under conditions of secondary electron equilibrium and neglecting bremsstrahlung production the *air kerma* and *absorbed dose* to air have identical values.

Kerma-area product (KAP)

Product of the air kerma and the cross-sectional area of the beam perpendicular to the beam axis, both quantities being measured at the same distance from the focal spot. Sometimes the term air kerma area product is used. The unit of dose area product is Gy m².

Laser

A source of an intense, coherent, directional beam of optical radiation. Also, the acronym for: Light Amplification by Stimulated Emission of Radiation. A laser usually is composed of an energy source, a resonant cavity, and an active lasing medium.

Lateral profile

Variation of the dose $D(x)$, where x is a spatial coordinate on a straight line going perpendicularly through the axis of a radiation field.

Lateral spread

Lateral dose contributions beyond the geometrical borders of radiation fields due to scatter.

Leakage current

Total detector current flowing at the operating bias in the absence of radiation.

Linear accelerator

A type of particle accelerator that is overwhelmingly used in radiation oncology today to produce high-energy photon and electron beams, see also *Cyclotron* and *Synchrotron*.

Linear attenuation law

In a beam of monoenergetic photons (or other indirectly ionizing particles) crossing a layer of material the exponential decay of the fluence of particles which have not undergone any interaction.

Linear attenuation coefficient, μ

Quotient dN_p/N_p by ds , where N_p is the number of primary photons impinging on the attenuating layer of thickness ds and dN_p the number of photons undergoing at least one interaction

$$\mu = \frac{1}{N_p} \frac{dN_p}{ds} .$$

Linear collision stopping power

That part of the linear stopping power resulting from collisions, i.e. by excitations and ionizations and not from the production of bremsstrahlung. This term is also referred to as collision stopping power.

Linear Energy Transfer (LET)

Describes a radiation quality in regard to its energy deposition per unit track length, see also *Densely ionizing* and *Sparsely ionizing*.

Linear radiative stopping power

That part of the linear stopping power caused by processes of bremsstrahlung production and not from collisions, i.e., by excitations and ionizations. This term is also referred to as radiative stopping power.

Linear stopping power

Quotient dE over ds , where dE is the average energy lost in single or multiple scattering processes by charged particles of energy E in a certain material after propagating the distance ds

$$s = \frac{dE}{ds} .$$

This term is also referred to as stopping power.

Linear-non-threshold (LNT) dose response model

Practical system of radiological protection recommended by ICRP based on the assumption that at doses below about 100 mSv a given increment in dose will produce a directly proportionate increment in the probability of incurring cancer or heritable effects attributable of radiation.

Lymphocyte

White blood cells produced in lymphoid tissue that initiate adaptive, antigen-specific immune responses. Some T-lymphocytes are cytotoxic; B-lymphocytes secrete antibodies.

 μ -map

A 3D- or 2D transverse image of the photon attenuation factor obtained via transmission imaging, usually utilising computed tomography technique. A μ -map is the basis for attenuation corrections of the SPECT-images.

Magnetic field strength, H

A field vector, H , that is equal to the *magnetic flux density* divided by the *magnetic permeability* of the medium. Expressed in units of ampere per meter (A m^{-1}).

Magnetic flux density, B

The force on a moving unit positive charge at a point in a magnetic field per unit velocity. A vector field quantity, B , expressed in tesla (T). The magnetic flux density is often called the magnetic field, because the two terms differ only by a proportionality factor in vacuum.

Magnetic permeability

The scalar or vector quantity which, when multiplied by the magnetic field strength, yields magnetic flux density; expressed in unit henry per meter (H m^{-1}). *Note:* For isotropic media, magnetic permeability is a scalar; for anisotropic media, it is a tensor quantity. Synonym: absolute permeability. If the permeability of a material or medium is divided by the permeability of vacuum (magnetic constant) μ_0 , the result is termed relative permeability (μ).

Magnetic resonance imaging (MRI)

A diagnostic imaging technology that exploits the tendency of nuclei with magnetic moments (typically protons) to precess about static magnetic fields at frequencies proportional to the local value of the static magnetic field. Resonant radiofrequency fields excite these nuclei, effectively reporting their locations. Magnetic field gradients are used to spatially encode the region of interest.

Magnetic resonance spectroscopy (MRS)

MRS uses principals similar to those used in MRI. However, uniform static magnetic fields are used without magnetic field gradients in MRS. The molecular environment near the relaxing nuclei cause differences in received frequencies. Frequency components of the resulting spectra provide a chemical analysis of the region of interest. MRS may be done *in vivo* in MR scanners. Most MRS is done *in vitro* on machines that accept only small chemical samples.

mAs

Product of the exposure time and the X-ray tube current.

Mass-energy absorption coefficient

Mass-energy transfer coefficient multiplied by the factor $(1-g)$, where g is the *bremsstrahlung* yield in the matter concerned.

Mass-attenuation coefficient

Linear attenuation coefficient divided by the density of the attenuating material.

Mass-energy transfer coefficient

Describes the energy transfer of photons penetrating matter from the photons' point of view (see also *Mass-energy absorption coefficient*). Quotient of $dE_{\text{tr}}/(\rho EN)$ and ds , where E and N are the energy and number of the incoming photons, respectively, dE_{tr} is the average kinetic energy transferred from photons to charged particles when the photons travel the distance ds , and ρ is the density of the material.

Mass-stopping power

Linear stopping power divided by the density of the material.

Mathematical phantom

The description of a reference person and its organs and tissues by geometric figures like ellipsoids. It is used for the calculation of S values for penetrating radiation.

Mean (average) glandular dose

The mean absorbed dose in the glandular tissue (excluding skin) in a uniformly compressed breast of, for example, 50% adipose, 50% glandular tissue composition. The reference breast thickness should be specified.

Mean range

Concept for charged particles penetrating matter to describe the depth where the particle comes to rest.

Measuring assembly

A device for measuring the current or charge from the detector and converting it into a form suitable for display, control or storage.

Median threshold

The threshold value within a statistical distribution at which 50% of subjects have greater thresholds and 50% have lesser thresholds.

Medical exposure

Exposure incurred by patients as part of their own medical or dental diagnosis or treatment; by persons, other than those occupationally exposed, knowingly, while voluntarily helping in the support and comfort of patients; and by volunteers in a programme of biomedical research involving their exposure.

Metabolism

The biochemical reactions by which energy is made available for the use an organism from the time a nutrient substance enters, until it has been utilized and the waste products eliminated.

Microdosimetry

See *Small scale dosimetry*.

Microwaves

Electromagnetic radiation with the frequency range of 300 MHz – 300 GHz.

MIRD, Medical Internal Radiation Dose Committee

A committee of the US Society of Nuclear Medicine. It develops and recommends models for internal dose assessment in nuclear medicine.

MIRDOSE

A computer program for calculating internal radiation dose estimates for radionuclides used in nuclear medicine. It now has been substituted by OLINDA/EXM.

Model equation

Equation by means of which the indicated value(s) of a measuring instrument are converted to the value of the quantity to be measured.

Monitor

An instrument used on electron accelerators and X-ray tubes to survey the stability of the output (rates of absorbed dose to water or air kerma) during irradiation or to compare values of absorbed dose to water or air kerma in successive irradiations.

Monte Carlo calculation

Numerical simulation of particle histories crossing space and matter with the objective of calculating energy and directional distributions of primary and secondary radiation. From the properties of the radiation field dose distributions may be calculated taking into account the geometrical conditions and stochastic variation of interaction with matter. Correct distributions are obtained by sampling large

numbers of particle histories, e.g. of the order of 10^9 . This method is also referred to as Monte Carlo simulation or briefly as MC simulation.

Mutagen

A substance that is able to cause a mutation.

Mutation

Any detectable and heritable change in the DNA at a specific site in the genome of a cell by an agent (mutagen) such as ionizing radiation.

Narrow beam geometry

A beam model where scattered radiation is assumed to be completely removed from the beam.

Neoplastic initiation of cancer

Irreversible cellular damage which provides the potential in cells for the development of cancer, resulting from damage to DNA and leading to gene mutations in single target cells in tissues.

Neoplastic promotion of cells

Cells irreversibly damaged by irradiation receive an abnormal growth stimulus and begin to proliferate in a semi-independent manner.

Neural network

Group of interacting neurons.

Neuron

Nerve cell, specialised for the transmission of neural information.

NIS

The sodium-iodide symporter (NIS) is an ion pump that actively transports iodide (I^-) into thyroid epithelial cells.

Non-invasive high-voltage measuring instrument

Device placed in the X-ray beam which allows the determination of the X-ray tube voltage through (simple) spectroscopic methods.

Non-ionizing radiation (NIR)

Includes all radiations and fields of the electromagnetic spectrum that do not normally have sufficient energy to produce ionization in matter; characterized by energy per photon less than about 12 eV, which is equivalent to wave lengths greater than 100 nm, or frequencies lower than $3 \cdot 10^{15}$ Hz. From a pragmatic point of view, magnetostatic and electrostatic fields are also dealt within the framework of NIR. The field of NIR comprises also pressure waves such as ultrasound and infrasound including airborne ultrasound and infrasound. That is on either side of the audible frequency range (20 Hz-20 kHz).

Normal operating mode, MRI

Because of the uncertainty over identified deleterious effects during MRI procedures, it is recommended that exposure limits for the patients be divided into three tiers. The normal operating mode is used for routine MR examinations for all patients.

Occupational exposure, EMF

Exposure experienced by adults who are generally exposed under known conditions and are trained to be aware of potential risk and to take appropriate precautions.

Occupational exposure, Ionizing radiation

Exposure of workers, for example of medical staff in nuclear medicine or radiology departments.

OLINDA/EXM, Organ Level INternal Dose ASsessment/EXponential Modeling

A computer program for calculating internal radiation dose estimates for radionuclides updating MIRDose.

Operational quantities

Quantities, such as the *ambient dose equivalent* and the *personal dose equivalent*, used in practical applications for monitoring and investigating situations involving external exposure. They are defined for measurements and assessment of doses in the body. In internal dosimetry, no operational dose quantities have been defined which directly provide an assessment of *equivalent* or *effective dose*.

Optical density

The degree of blackening of processed X-ray or photographic film. Numerically equal to the decadal logarithm of the ratio of light incident on the film to that transmitted through the film.

Optical radiation

Electromagnetic radiation comprising ultraviolet, visible and infrared radiations.

Optimisation (of protection and safety)

The process of determining what level of protection and safety makes exposures, and the probability and magnitude of accidental exposures, as low as reasonably achievable, economic and societal factors being taken into account. In the context of medical exposures of patients, managing patient doses to be commensurate with the medical purpose.

Organ/tissue dose (Gy)

The average absorbed dose in a specified organ or tissue.

Pair production

Conversion of a photon into an electron-positron pair. The minimum photon energy corresponds to the total rest mass of the electron-positron pair.

Palliative

Aiming at preventing or relieving the suffering from the disease.

Parallel-plate chamber

An ionization chamber with a measuring volume of between 0.01 cm^3 and 0.5 cm^3 bounded by parallel electrodes. The chamber is intended to be used with the electrodes perpendicular to the axis of the radiation beam.

Partial-volume effect (PVE)

The term refers to two distinct phenomena that make intensity values in images differ from what they ideally should be. The first effect is the 3-dimensional (3D) image blurring introduced by the finite spatial resolution of the imaging system. The spatial resolution is limited by the detector design and by the reconstruction process. The resulting 3D blurring causes spillover between regions. Because of the finite spatial resolution, the image of a small source is a larger but dimmer source. Part of the signal from the source “spills out” and hence is seen outside the actual source. The second phenomenon causing PVE is image sampling. The radiotracer distribution is sampled on a voxel grid. Obviously, the contours of the voxels do not match the actual contours of the tracer distribution. Most voxels therefore include different types of tissues. This phenomenon is often called the tissue fraction effect. The signal intensity in each voxel is the mean of the signal intensities of the underlying tissues included in that voxel. Even if the imaging system had perfect spatial resolution, there would still be some PVE because of image sampling. Ideally, compensation for PVE should account for both the finite resolution effect and the tissue fraction effect. Motion, especially respiratory motion, also introduces a blurring effect that results in additional PVE.

Particle fluence

See *Fluence*.

Patient dose (exposure)

Generic term used for a variety of quantities applied to a patient or group of patients. The quantities are related and include *absorbed dose*, *incident air kerma*, *entrance surface air kerma*, etc.

Pencil Beam

Radiation beam of infinitesimally small cross-section used as a mathematical concept for dose calculations with limited accuracy for inhomogeneous media, see also *Collapsed Cone*.

Peptide

A short polymer of amino acids linked by peptide bonds. The term peptides includes the term proteins. There are small (oligopeptides) and large (polypeptides, like proteins) peptides.

Percentage depth dose (PDD)

Ratio of the values of the absorbed dose to water on the axis of the beam at a certain depth inside a phantom to that at the dose maximum. Often a source-to-phantom distance of 100 cm, a field size of 10 cm × 10 cm and a depth of 10 cm is chosen, in which case this quantity is referred to as PDD(10).

Peripheral nerve

Nerve found outside the central nervous system and leading to and from the central nervous system. The cell bodies lie within the spinal cord, but the peripheral nerves (axons) terminate on muscle fibres or in specialised sensory receptors throughout the body.

Permanent filtration

See *Inherent filtration*.

Permeability

See *Magnetic permeability*.

Permittivity, dielectric constant

A constant defining the influence of an isotropic medium on the forces of attraction or repulsion between electrified bodies. Symbol: ϵ . Unit: farad per meter ($F\ m^{-1}$).

Permittivity, relative

The *permittivity* of a material or medium divided by the *permittivity* of the vacuum, ϵ_0 .

Personal dose equivalent, $H_p(d)$

An *operational quantity*: the dose equivalent in soft tissue (commonly interpreted as the ‘ICRU sphere’) at an appropriate depth, d , below a specified point on the human body. The unit of personal dose equivalent is joule per kilogram ($J\ kg^{-1}$) and its special name is sievert (Sv). The specified point is usually given by the position where the individual’s dosimeter is worn.

Perturbation factor

Numerical factor by means of which the response of a real detector is converted to the response of an ideal Bragg-Gray detector.

PET

A nuclear medicine imaging technique which produces a three-dimensional image of functional processes in the body. The system detects pairs of 511 keV gamma rays emitted indirectly by a positron-emitting radionuclide (tracer), which is introduced into the body on a biologically active molecule. Three dimensional imaging is often accomplished with the aid of a CT during the same session, in the same machine.

PET/CT

A multi-modal imaging equipment where *positron emission tomography* and *computed tomography* are combined such that the two types of image are obtained during the same session and in the same machine. Several other kinds of multi-modal imaging machines are also available.

Phantom

An object used to absorb and/or scatter radiation in a way equivalent to that by a patient and hence to aid estimation of radiation doses and test imaging systems without actually exposing a patient. It may be an anthropomorphic or a physical test object.

Phosphene

The perception of flickering light in the periphery of the visual field that occur in response to magnetic fields or by direct electrostimulation. The effect is believed to result from the interaction of the induced current with electrically excitable cells in the retina.

Photoelectric effect

Interaction in which the total energy of an incoming photon is transferred to an electron of an atomic shell. The electron is ejected, the ionized atom reaches its ground state by the emission of a characteristic X-ray photon or through an Auger process.

Photonuclear reactions

Type of photon interaction with the atomic nucleus.

Pixel

Picture element, the smallest component of a digital image. Depending on its position in the image a pixel represents a specific location of the imaged object. The pixel value represents the number of registrations that may be attributed that location.

Placenta

An organ responsible for the exchange of nutritive substances, oxygen, and waste products between the foetus and the mother's blood supply. The exchange takes place via diffusion through a membrane, the so called placenta barrier.

Planar fluence

Quotient of dN by da , where dN is the number of particles incident on a *plane* of cross-sectional area da , $\Phi = dN/da$, unit: m^{-2} .

Plane wave

An electromagnetic wave in which the electric and magnetic field vectors lie in a plane perpendicular to the direction of wave propagation, and the magnetic field strength (multiplied by the impedance of space) and the electric field strength are equal.

Plank's constant

$h = 6.626\ 068\ 96 \times 10^{-24}\ Js$.

Planning Target Volume (PTV)

Tissue volume encompassing the tumor in a way to allow for microscopic extensions of the primary tumor plus some 'safety' margin to account for geometrical uncertainties. It is usually defined by the radiotherapist during treatment planning.

Point of measurement

Point at which the dose is to be determined.

Polarity effect

Variation of the response of an ionization chamber on polarity of the bias voltage.

Polycytemia

A disease characterized by an abnormal high proportion of red blood cells in the blood. One possible reason might be a too high production rate of red blood cells in the red marrow.

Polymethylmethacrylate (PMMA)

A polymer plastic commercially available as Perspex or Lucite.

Portal imaging

The formation of a radiological image by means of the radiation transmitted through the patient during external radiation therapy.

Positron emission tomography (PET)

See *PET*.

Potential difference

The same term as voltage. This term arises because the voltage is the potential to do work.

Power density

Power crossing a unit area normal to the direction of wave propagation. Unit: watt per square meter (W m^{-2}).

Power frequency

In Europe the frequency of the mains (and the rest of the electricity system) is 50 Hz, in US it is 60 Hz. "Power frequencies" is often used to cover both 50 Hz or 60 Hz and the first few harmonics.

Practical peak voltage, \hat{U}

Weighted mean of the X-ray tube potential according to:

$$\hat{U} = \frac{\int_{U_{\min}}^{U_{\max}} p(U)w(U)U dU}{\int_{U_{\min}}^{U_{\max}} p(U)w(U) dU},$$

where $p(U)$ is the distribution function for the voltage, U , and $w(U)$ is a weighting function, U_{\max} is the highest voltage in the interval and U_{\min} is the lowest voltage in the interval. The unit of the quantity practical peak voltage is the volt (V). The weighting function, $w(U)$, is determined in such a way that the practical peak voltage is that constant tube voltage which produces the same air kerma contrast behind a specified phantom as the non-DC voltage under test.

Practical range, R_p

Beam specifier in electron dosimetry. It denotes the point of intersection of the tangent to the point of inflection on the falling part of the depth-dose curve and the linear extrapolation of the bremsstrahlung background to smaller depth values.

Primary radiation

Radiation emitted by the radiation source. The primary radiation is considered to comprise that part of scattered radiation generated within the source itself.

Primary standard

Device of highest metrological quality allowing the measurement of a quantity according to the definition of that quantity. With a primary standard the unit of a quantity is realized. An example is the measurement of the absorbed dose to water by means of a water calorimeter. The temperature rise generated by the radiation in the water divided by the specific heat capacity of water is equal to the energy imparted to a mass element of the water divided by the mass element.

Progression of cancer

Invasion of cancer cells into adjacent normal tissues, circulation of neoplastic cells in the blood and lymphatic systems, and the establishment of metastases at other sites of the body.

Protective sheath

Cap to be placed on an ionization chamber so that the ensemble of the non-water tight chamber and the cap represents a water tight unit with which measurements inside a water phantom can be conducted. It is important that a protective sheath does not interfere with the vented nature of an ionization chamber.

Public exposure

All exposure experienced by members of the general public, excluding occupational exposure and exposure during medical procedures.

Quality Assurance (QA)

The function of a management system that provides confidence that specified requirements will be fulfilled, comprising all those planned and systematic actions necessary to provide confidence that a structure, system or component will perform satisfactorily in service.

Quality Control (QC)

Part of *quality assurance* intended to verify that structures, systems and components correspond to predetermined requirements.

 R_{50}

Beam specifier in electron dosimetry. It denotes the depth in which the dose has fallen to half of its value in the dose maximum.

Radiance, L

Radiant flux (or radiant power) output per unit solid angle per unit area usually expressed in $\text{W sr}^{-1} \text{cm}^{-2}$.

Radiant energy, Q

Energy in the form of electromagnetic waves usually expressed in units of joules (W s).

Radiant exposure, H

The total energy per unit area upon a given surface in a given time interval, usually expressed in J cm^{-2} .

Radiant intensity, I

Radiant flux emitted from the source per unit solid angle (steradian), in the direction of propagation, expressed in W sr^{-1} .

Radiant power or radiant flux, Φ

The time rate of flow of radiant energy. Unit: watt (W).

Radiation effect, deterministic

A radiation effect for which generally a threshold level of dose exists above which the severity of the effect is greater for a higher dose (see *Deterministic biological effects*).

Radiation effect, stochastic

A radiation effect, generally occurring without a threshold level of dose, whose probability is proportional to the dose and whose severity is independent of the dose (see *Stochastic biological effects*).

Radiation induced mental retardation

Proper development and function of the human brain depends upon an elaborate sequence of neuronal cellular proliferation and migration within the cerebral cortex. Any disturbance of this sequence could lead to abnormality – called mental retardation – since the normal function of the nervous system depends upon the proper location of the neuronal cells. Radiation induced mental retardation was observed in children exposed to various radiation doses from atomic bombing in Hiroshima and Nagasaki at various developmental stages in utero.

Radiation quality

One or more parameters characterizing the quality of a radiation beam. In the case of X-rays the X-ray tube voltage and the half-value layer are frequently used. For high-energy photon radiation percentage depth dose or the radiation quality index and for electron beams the parameter R_{50} are frequently used.

Radiation weighting factor, w_R

The factor characterizes the biological effectiveness of a specific kind of ionizing radiation relative to photons. For external irradiation w_R is a body-averaged value representing a mean value for the relative biological effectiveness of all tissues of the body. Thereby, any local variation of the radiation quality in the human body which may result from the generation of secondary radiation of different types in the body is not explicitly considered.

Radiative mass-stopping power

Linear radiative stopping power divided by the density of the material.

Radionuclide generators

Allow the on-site production of purely gamma-emitting radioisotopes by chemically eluting (extracting) the daughter from a transient mother-daughter equilibrium.

Radioactive isotopes, radioisotopes

Radioactive member(s) of an isotope family, e.g. ^{131}I .

Radioactivity

The phenomenon whereby atoms undergo spontaneous random disintegration, usually accompanied by the emission of ionizing radiation.

Radiochromic film

Film whose optical density is increased by being exposed to ionizing radiation. It consists of a sensitive layer sandwiched between two coatings which polymerizes when exposed to radiation. The film is also referred to Gaf-chromic film.

Radiofrequency (RF)

Any frequency at which electromagnetic radiation is useful for telecommunication, usually the frequency range 300 Hz – 300 GHz.

Radiopharmaceutical

A radioactive compound used to diagnose certain medical problems or treat certain diseases. They may be given to the patient by injection, by ingestion or by inhalation.

Rayleigh scattering

Photon interaction in which the photon undergoes a change in the direction of propagation without experiencing a net exchange of energy.

Recombination correction

Procedure to account for that part of charge carriers initially generated by ionizing radiation having undergone a recombination process.

Red bone marrow

The component of marrow which contains the bulk of the haematopoietic stem cells.

Reduction factor, EMF

Reduction of the effect threshold to compensate for various sources of uncertainty in the guideline setting process.

Reference air kerma rate (brachytherapy)

Air kerma rate at a distance of 1 m from the source in a direction perpendicular to the source axis.

Reference conditions

Conditions under which all influence quantities and instrument parameters have their reference values. Under reference conditions the calibration factor of a dosimeter is applicable without further corrections. The reference conditions should be stated in the calibration certificate.

Reference depth

Depth inside a phantom for which the radiation quality dependent correction factor (k_Q -factor) has been determined. A depth different from the reference depth the uncertainty of the k_Q -factor tends to be larger than at the reference depth.

Reference level, EMF

Exposure level provided for practical exposure assessment purposes to determine whether the basic restrictions are likely to be exceeded. Some reference levels are derived from relevant basic restrictions using measurement and/or computational techniques and some address perception and adverse indirect effects of exposure to EMF.

Reference person

The definition of average persons of different age and sex for radiation protection purposes. They have been defined in ICRP Publication 89, 2002.

Reference point of a detector

The point of the detector by means of which the detector is positioned in the radiation field. The reference point should be marked on the detector assembly by the manufacturer of the instrument. If this proves impossible, the reference point should be indicated in the accompanying documentation supplied with the instrument.

Reference value

Parameter values for biokinetic and dosimetric models to calculate doses for reference persons.

Relative biological effectiveness (RBE)

Biological effectiveness of a specific kind of ionizing radiation relative to that of Co-60 photons.

Relative radiation detriment

Detriment is a measure of total harm that could eventually be experienced by an exposed group of individuals and its descendants as the result of the group's exposure to an ionizing radiation source. Health detriment is included as part of the total detriment. The definition of detriment by the ICRP is the expected number of cases of a radiation induced health effect weighted by a factor representing the severity of the effect. This weighting factor is taken as 1 for the death of individuals and for severe hereditary effects in descendants.

Response

The ratio between the indication of the measuring assembly and the conventional true value of the measured quantity at the position of the reference point in space. The response under reference conditions is the inverse of the calibration factor.

Restricted stopping power

Quotient dE_Δ over ds , where dE_Δ is the average energy lost by charged particles of energy E in a certain material after propagating the distance where the kinetic energy of the particle liberated in the ionization process does not exceed the energy Δ :

$$S_\Delta = \frac{dE_\Delta}{ds} .$$

ROI, region of interest

A region in a digital image may be virtually marked for special readout of mathematical operations.

Root mean square (rms)

Certain electrical effects are proportional to the square root of the mean of the square of a periodic function (over one period). This value is known as the effective or root-mean-square (rms) value, since it is derived by first squaring the function, determining the mean value of the squares obtained, and taking the square root of that mean value. Often used for averaging the magnitude of time-varying electric and magnetic fields.

Safety

The achievement of proper operating conditions, prevention of accidents, or mitigation of accident consequences.

Safety assessment

The process, and the result, of analysing systematically and evaluating the hazards associated with radiation sources and practices, and associated protection and *safety* measures. Comprises an analysis to predict the performance of an overall system and its impact, where the performance measure is the radiological impact or some other global measure of the impact on *safety*; also, the systematic process that is carried out throughout the design process to ensure that all the relevant *safety* requirements are met by the proposed (or actual) design.

Safety factor, EMF

A reduction factor used in deriving Basic Restrictions and Reference Levels which provides for the protection of exceptionally sensitive individuals, uncertainties in the determination of a threshold level of exposure, uncertainties concerning threshold effects due to pathological conditions or drug treatment, and uncertainties in induction models.

Saturation

The absence of recombination of charge carriers of opposite sign in ionometry. In reality this cannot be achieved completely. Therefore a correction is applied by means of which the amount of charges lost by recombination is determined.

Scattering

Group of interactions in which the incoming particle experiences a change in its direction of propagation. A scattering process may but does not need to be accompanied by a transfer of energy from the incoming particle to the interaction partner. In the absence of a net exchange of energy one speaks of coherent scattering.

Scattering power

Describes the deflection of charged particle radiation from its incident path.

Scattered radiation

Radiation which, through interaction processes, has changed its direction and in most cases its energy. In diagnostic radiology scattered radiation does not contain image information.

Secondary electron equilibrium (SEE)

Idealized state of a photon/electron radiation field in which the amount of energy deposited in the volume of interest by electrons coming from outside is compensated by the energy carried out of the volume of interest by electrons set free in the volume of interest. More rigorously one postulates that not only the two amounts of energy are in mutual balance but that the equilibrium is created by two groups of electron interactions with the same spectral fluence distribution.

Secondary standard dosimeter

A dosimeter calibrated directly against a primary standard.

Security

The prevention and detection of, and response to, theft, sabotage, unauthorised access, illegal transfer, or other malicious acts involving nuclear material, other radioactive substances, or their associated installations.

Selection bias

Bias resulting from a faulty way to select subjects for a study. Epidemiological studies depend on a reliable comparison between subjects with a disease and a reference population as to their exposure. If the subjects chosen for a study are not representative of the corresponding population, the comparison becomes flawed and the association between disease and exposure becomes biased.

Sievert

Unit of effective dose and equivalent dose, in J kg^{-1} .

Small scale dosimetry

The internal dosimetry could be subdivided into three levels related to the dose distribution within organs and tissues; macroscopic (The activity distribution is assumed to be uniform and the calculated mean absorbed dose serves as a good representation of the biological effect since the volumes are large compared to the range of emitted particles), small-scale (the mean absorbed dose serves as a poorer representation of the biological effect) and microscopic dosimetry (the smaller a volume, the larger stochastic effects are seen).

Source region

A region within the body in which a radioactive substance accumulates. Source regions are specified in biokinetic models.

Source strength

Generic term addressing the output of a radiation source in brachytherapy.

Spark discharge

The transfer of current through an air gap requiring a voltage high enough to ionize the air, as opposed to direct contact with a source.

Sparingly ionizing radiation

Categorizes a radiation quality in regard to its energy deposition per unit length. A rather arbitrary limit at 3–5 keV/ μm separates sparingly ionizing (energy transfer below 3–5 keV/ μm) from densely ionizing radiation.

Spatial average

The root mean square of the field over an area equivalent to the vertical cross section of the adult human body, as applied to the measurement of electric or magnetic fields in the assessment of whole-body exposure.

Spatial peak pulse average intensity, I_{SPPA} , Ultrasound

The spatial peak pulse average intensity is the maximum intensity in the beam averaged over the pulse duration (for pulses of non-constant amplitude)

Spatial peak temporal average intensity, I_{SPTA} , Ultrasound

The spatial peak temporal average intensity is the maximum intensity in the beam averaged over the pulse repetition period. I_{SPTA} is the best measure of the amount of heat delivered to a tissue by ultrasound. In diagnostic imaging, I_{SPTA} is usually below 100 mW cm^{-2} .

Specific effective energy, $\text{SEE}(\mathbf{r}_k \leftarrow \mathbf{r}_h)$

The equivalent dose in the target tissue r_k due to a nuclear transformation in the source tissue r_h .

Specific energy absorption

The energy absorbed per unit mass of biological tissues, (SA) expressed in joule per kilogram (J kg^{-1}); specific energy absorption is the time integral of the specific energy absorption rate.

Specific energy absorption rate (SAR)

The rate at which energy is absorbed in body tissues, in watt per kilogram (W kg^{-1}); SAR is the dosimetric measure that has been widely adopted at frequencies above about 100 kHz.

Specific gamma (ray) constant, Γ

A specific gamma ray dose constant is a quantity for correlating the dose-equivalent rate (per unit of activity) for a radionuclide at a specified distance. For example, Γ (in tissue) for a point source containing ^{137}Cs at a radial distance of 1 meter of air is $1 \cdot 10^{-4} \text{ mSv h}^{-1}$ per MBq.

SPECT

Single photon emission computed tomography. A technique using rotating gamma cameras for reconstructing transversal nuclear medicine images.

SPECT/CT

A gammacamera equipped with an X-ray tube for CT (X-ray computed tomography) imaging. The CT- and gammacamera images are fused for improved diagnostics. The CT images may also be used for producing an attenuation map (μ -map) that may be used for attenuation correction of the gammacamera images.

Spread-out Bragg peak

By modulating the energy of a beam of protons or heavier ions with a certain amplitude the position of the Bragg-peak can be made to cover a certain range of depths inside a phantom.

Stakeholder

Interested party; concerned party. The term stakeholder is used in a broad sense to mean a person or group having an interest in the performance of an organisation.

Standard

- (1) A documented agreement containing technical specifications or other precise criteria to be used consistently as rules, guidelines or definitions of characteristics to ensure that materials, products, processes and services are fit for their purpose.
- (2) A legally enforceable limit for a substance or an agent intended to protect human health or the environment. Exceeding the standard could result in unacceptable harm.

Static field

A field vector that does not vary with time.

Steradian

The unit of measure for a solid angle. There are four steradians in a sphere.

Stochastic biological effects

The probability of the occurrence of an biological effect induced by ionizing radiation, but not its severity, depending on the radiation dose. This proves true for somatic and hereditary effects, which may start from a cell modified by interaction with ionizing radiation. The energy transfer from incident particles of ionizing radiation to the biological target material is a stochastic process, too. Malignant disease and heritable effects for which the probability of an effect occurring, but not its severity, is regarded as a function of dose without threshold.

Stopping power

See *Linear stopping power*.

Straggling

Term used for describing radiation transport phenomena of charged particles. It denotes the processes by means of which the particles of an initially monoenergetic and monodirectional beam (of identical particles) experience a spread in energy (energy straggling) and in the direction of propagation (directional straggling).

S-value, $S(r_k \leftarrow r_h)$

The absorbed dose in the target tissue r_k due to a nuclear transformation in the source tissue r_h .

Synchrotron

A type of particle accelerator used in radiation oncology to produce heavy charged particle beams, see also *Cyclotron* and *Linear accelerator*.

Target tissue

A radiosensitive tissue within the body for which the tissue dose is calculated.

Teletherapy

Form of radiation therapy in which the radiation source is outside the body. This form of treatment is also referred to as external radiation therapy.

Tenth value layer (TVL)

The thickness of a specified material required to attenuate a beam of radiation to one-tenth of its original intensity.

Teratogen

An agent that can cause birth defects.

Thermoluminescent dosimeter (TLD)

A type of radiation dosimeter. A TLD measures ionizing radiation exposure by measuring the amount of visible light emitted from a crystal in the detector when the crystal is heated. The amount of light emitted is dependent upon the radiation exposure.

Thermoluminescent dosimetry (TLD)

Form of solid state dosimetry in which metastable electronic interband levels in the TLD material are populated by means of ionizing radiation. Heating the exposed TLD detectors results in the emission of light which is recorded by means of a suitable detector, e.g., a photo-multiplier tube.

Thimble type chamber

Ionization chamber in which the outer electrode takes the form of a rigid cylindrical wall closed at one end and mounted at the other on the supporting stem. This type of chamber is also referred to as compact chamber.

Threshold

The lowest dose of an agent at which a specified measurable effect is observed and below which it is not observed.

Threshold dose

In radiological protection cautious approximation to the tolerance dose, which is defined as the amount of radiation received during conventional treatment below which unacceptable effects do not occur in more than in few percent of patients within 5 years following radiation treatment.

Thyroid stunning

A temporary reduction in the ability of normal thyroid tissue or differentiated thyroid cancer to trap or retain a therapeutic activity of ^{131}I -iodide following a prior administration of a diagnostic activity of ^{131}I - (or ^{123}I -) iodide.

Time-integrated activity, $\tilde{A}(i, T_D)$

The activity in a compartment i integrated up to time T_D . In nuclear medicine in general T_D is taken to be infinity. In the unit s the time-integrated activity equals the number of nuclear transformations in i .

Tissue weighting factor, w_T

Since the relationship between the probability of stochastic effects and equivalent dose is found to depend on the organ or tissue irradiated, ICRP has defined the quantity w_T , which represents the relative contribution of that organ or tissue to the total of the stochastic effects. A dimensionless factor used to weight the equivalent dose in a tissue or organ.

Tissue-equivalent material

Material which absorbs and scatters a specified ionizing radiation to the same degree as a particular biological tissue.

Tissue-maximum ratio (TMR)

Special case of the tissue-phantom ratio where z_1 is chosen to coincide with the depth of the dose maximum, z_{\max} .

Tissue-phantom ratio (TPR)

Beam specifier for high-energy photon radiation. It denotes the ratio of doses in depths z_2 and z_1 inside a phantom at constant source-to-detector distance (SDD):

$$\text{TPR} = \frac{D(z_2)}{D(z_1)} \text{ with } z_2 > z_1 .$$

The TPR is essentially independent of the beam divergence.

Total filtration

The total of inherent filtration built into the X-ray tube and added filtration.

TPR_{20,10}

Special case of the tissue-phantom ratio with $z_1 = 10$ cm and $z_2 = 20$ cm and a field size of 10 cm \times 10 cm on the phantom surface.

Trabecular bone

In the ICRP models trabecular bone are those bones with a surface/volume ratio larger than 60 cm⁻¹.

Traceability

See *Traceable calibration*.

Traceable calibration

Calibration traceable to a primary standard through an unbroken chain of calibrations.

Transients

Brief bursts of high frequency fields, usually resulting from mechanical switching of AC electricity.

Treatment Planning (Systems)

Computer simulation of a radiation oncology treatment course before real application. Eventually positive evaluation of the results leads to the application of the simulation (irradiation) parameters to the patient.

Tube current-exposure time product (mAs)

The product of X-ray tube current (mA) and the exposure time in seconds (s).

Tube loading

The tube current-exposure time product that applies during a particular exposure.

Ultraviolet radiation

Electromagnetic radiation found between X-rays and light in the electromagnetic spectrum. Has subregions UVA, UVB, UVC.

Unperturbed field

The electric or magnetic field, generated by a source, that is uninfluenced by the presence of conducting objects, including the human body, or sections of it.

Unrestricted stopping power

See *Linear stopping power*.

Uptake

The fraction of a substance in the systemic circulation which is taken up by a specified organ.

Visible radiation

Electromagnetic radiation capable of producing the sensation of vision and found between ultraviolet and infrared radiations in the electromagnetic spectrum.

Voltage-gated ion channel

Cell membrane proteins that allow the passage of particular ion species across the cell membrane in response to the opening of a molecular “gate” which is steeply sensitive to the transmembrane voltage. They are associated with electrical excitability.

Volume CTDI, CTDI_{vol}

Weighted CTDI corrected for the CT pitch factor:

$$\text{CTDI}_{\text{vol}, \text{H}} = \frac{\text{CTDI}_{\text{w}, \text{H}}}{p} \quad \text{or} \quad \text{CTDI}_{\text{vol}, \text{B}} = \frac{\text{CTDI}_{\text{w}, \text{B}}}{p} .$$

Voxel

A three-dimensional computational element with sides of 1-10 mm used to represent animal and human tissues in dosimetry models.

Voxel phantom

The definition of a specified person based on medical images for the calculation of S values. There are also modifications of such models to describe reference persons.

Water-equivalent thickness

Thickness of a water layer which produces at the point of measurement the same dose as a given layer of a different material.

Waveform

A single component of the field measured as a function of time by an instrument with a response time much faster than the field’s frequency of oscillation. The term also refers to the shape of the wave as displayed on a graph or oscilloscope trace.

Wavelength, λ

The distance between two successive points of a periodic wave in the direction of propagation, in which the oscillation has the same phase. Symbol: λ . Unit: meter (m).

Weighted CTDI_{w,H} or CTDI_{w,B}

Weighted mean of the air kerma in the phantom cross-section for one scanner rotation. The average is obtained from the CTDI-values in the centre (CTDI_{100,c}) and on the periphery (CTDI_{100,p}) of the head or CT head or body phantom.

$$\text{CTDI}_{w,H} = (1/3) \text{CTDI}_{100,H,c} + (2/3) \text{CTDI}_{100,H,p}$$

$$\text{CTDI}_{w,B} = (1/3) \text{CTDI}_{100,B,c} + (2/3) \text{CTDI}_{100,B,p}$$

CTDI_{w,H} and CTDI_{w,B} are usually given in units of mGy. w_T are the *tissue weighting factors* characterising the relative sensitivity of the various tissues with respect to stochastic effects resulting from ionizing radiation exposure and H_T is the equivalent dose in one of the 13 specified tissues and organs.

 W -value

Average energy required to produce an ion pair in dry air; for electrons with energies above 5 keV: $W = 33.97$ eV.

X-ray tube current

Electron current flowing inside the X-ray tube from the cathode to the anode.

X-ray tube voltage ripple, r

The ratio, expressed as a percentage, defined for a given current by the formula:

$$r = \frac{U_{\max} - U_{\min}}{U_{\max}} 100\%$$

where U_{\max} is the maximum value and U_{\min} the minimum value between which the voltage oscillates.

X-ray tube voltage

Potential difference applied to an X-ray tube between the anode and the cathode. Usually, the X-ray tube voltage is given in kilovolts (kV).

X-ray unit

Assembly comprising a high voltage supply, an X-ray tube with its protective housing and high voltage electrical connections.

 z_{\max}

Beam specifier; it denotes the depth in the phantom where the dose maximum on the beam axis is located.