Fundamentals of Medical Physics in Diagnostics and Radiation Therapy

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Abstract

Medical physics can be defined as the application of physics to medicine in diagnostics and radiation therapy. The functions of a medical physicist in diagnostics include assisting in the operation of imaging systems, establishment of quality assurance programs, and the implementation of safety and accreditation standards. In radiation therapy, medical physicists contribute to the safety of the treatment as part of the treatment planning team and support the construction and use of machine systems, patient treatment planning, and quality assurance.

In accord with the principles of evidence-based medicine, clinical applications of medical physics are regulated and overseen by accredited independent professional bodies, such as the American College of Radiology and the Radiological Society of North America. By helping to deliver radiation doses that are as low as reasonably achievable (the ALARA principle), medical physicists seek to optimize imaging performance in radiology and radiation therapy. Consequently, the concepts and aspects of diagnostic and therapeutic physics are steadily integrated into successive bodies of the international medical physics report series as those disciplines develop.

Chapter - 1

Introduction to Medical Physics

Medical physics is a diverse discipline encompassing the application of physics concepts, theories, and principles in medicine. At its core, medical physics facilitates the safe and effective use of differing modalities used in diagnostics and therapy. Medical physicists working in diagnostic radiology undertake tasks related to the safe application of X-rays and other radiation-based techniques for image generation, including patients, operators, and the general public. Physiological responses to different external agents such as pressure, sound, light, and magnetic fields are exploited in other diagnostic imaging procedures, including ultrasound and MRI, providing highly useful images of human anatomy and functions. Medical physicists specializing in radiation therapy ensure safe and effective delivery of radiation dose to patients, affecting disease treatment, primarily cancer, while minimizing the unwanted side effects to surrounding organs and tissues.

Their primary goal is to protect the individual from the hazards of ionizing radiation while providing a level of exposure sufficient for the imaging or therapy task. The directives of various governing and legislative bodies have been adopted into practice by both the medical community and the patient and general public, resulting in three code-of-practice principles. These principles, often referred to as the ALARA (as low as reasonably achievable) principle of radiation protection, have been integrated into the work of the medical physicist and can be summarized as follows: the source of radiation should be used only when justified, the radiation exposure should be kept as low as possible, and the shielding should be provided to restrict exposure within the defined limits.

Modern practice is based on sound scientific principles supported by a wealth of clinical experience. Medical physics is governed by a stringent regulatory and ethical framework. Such regulation and professional application oversee the practice of all medical physics areas, covering external beam and brachytherapy, including radiological applications [1, 2, 3].

Branches of medical physics can be classified into two large areas: diagnostics and therapy. Diagnostic physics concerns the use of techniques that involve interaction of photons with the human body. An important aspect of all medical procedures is the necessity of weighing benefits against risks, for the patients and, ultimately, for public health. Medical practitioners are therefore required to minimize radiation dose while providing sufficient image quality for satisfactory diagnosis.

Medical Physics in diagnostics illustrates the fundamental physical principles behind each different modality of radiological imaging, from projection radiography and fluoroscopy, through surgery and computed tomography, to ultrasound and magnetic resonance techniques. Special attention is given to processes of radiation interaction with tissues, to image formation and encoded resolution, to dose estimation and monitoring, and to the technology of systems and sensors. Principles and techniques related to the study of radiation-induced effects in biological systems are also considered, such as growth of tumor cells and embryonic tissue, and determination of stochastic and deterministic risk functions [4, 5, 6].

Definition and scope of medical physics

Medical physics encompasses all aspects of physics that apply to medicine. Its long-term goal is to enhance the quality and safety of the use of radiation in the diagnosis and treatment of disease, as well as to imply physics-oriented improvements to other established or new medical imaging or therapeutic methods involving non-ionizing radiation. Ideally, it should illustrate how physics principles can improve patient care.

Medical physics practices in hospitals consist of diagnostics and therapeutics involving artificial beams or sources of ionizing radiation. Two primary functions are important for patient safety: ensuring the patient is exposed only to the appropriate amount of radiation for diagnosis or treatment and ascertaining that the imaging or therapeutic procedure is conducted optimally for the intended purpose. In diagnostic procedures, these functions can be quantified using image quality-related parameters, while in radiation therapy, they can be expressed through the concept of radiobiological modeling. The roles of a clinical medical physicist can be classified as a physicist directly involved in clinical operation and interpretation, a physicist contributing ideas to improve quality or safety, and a physicist supporting others in performing their roles safely and effectively [1, 7, 8, 1, 7, 8].

Historical evolution of diagnostic and therapeutic physics

The application of physics in medicine has a long history dating back to the discovery of X-rays and radioactivity. Long before the establishment of formal medical physics departments, scientists and engineers worked with medical practitioners to apply their knowledge of radiation and radioactivity to the diagnosis and treatment of disease. In the early 20th century, Thomas Edison, Ernest Rutherford, William Coolidge, and many others recognised the need to understand and quantify the effects of these early medical technologies and to develop techniques to refine the images or mitigate harmful effects. The establishment of formal medical physics departments and training programs allowed for the decline of ad-hoc and opportunistic solutions to medical problems, and ensured that formal error-reduction, exploratory experimentation, and the development of guidelines formed the basis of clinical medical physics activity as practised today.

The scientific method demanded that the hazards of radiation be quantitated, in order that equipment and techniques could be modified to reduce their incidence while maximising potential benefits and saving lives. With the application of simple dose-response relationships, it became possible to calculate the risk of fatal radiation-induced cancers (deterministic response) suffered by medical workers treating patients afflicted with serious illness, as well as by survivors exposed in the bombing of Japan. Further research demonstrated that specific cells of the tissues were responsible for deterministic effects, and that stochastic responses, such as radiation damage leading to cancers subsequent to radiation exposure, followed a different relationship. The knowledge gained enabled both workers on the bomb and subsequent patients receiving therapy to be reassured and the risk of exposure to be balanced against the benefit of therapy.

Roles of medical physicists

A medical physicist is a specialist trained to apply and promote the principles of physics in medicine. They are primarily involved in the areas of radiation therapy, diagnostic radiology, diagnostic nuclear medicine and, increasingly, ultrasound and magnetic resonance imaging. Medical physics is one of the few medical specialties for which training may be obtained outside of medicine and the medical licensure requirements.

Regulatory authorities in many countries require that each facility generating x- or gamma-ray radiation for diagnostic or therapeutic purposes employs its own medical physics service. The responsibilities of the medical physicist may include the following: general supervision of the generation of diagnostic and therapeutic radiation; design and construction of radiation shielding; evaluation of radiation detectors; performance of quality control tests; calibration of dosimeters used in the hospital; and consultation concerning the effect of radiation on patients undergoing diagnostic or therapeutic procedures. The International Atomic Energy Agency (IAEA; 2006) identified the absence of a clinical medical physicist as a deficiency that could contribute to radiation accidents and increased patient dose.

Medical physicists are required to be aware of the limits imposed by the available technology and physical processes. They must be able to assess the optimization of diagnostic and therapeutic techniques on the basis of appropriate objective criteria. These criteria include not only diagnostic and therapeutic efficacy but also patient risk. The United Nations Scientific Committee on the Effects of Atomic Radiation, the International Commission on Radiological Protection and national health and safety authorities maintain databases of radiation exposure worldwide. This is a requirement of the General Safety Requirements for Radiation Protection, which stipulate that radiation exposure incurred by patients and workers should be maintained at the lowest possible level consistent with the required physical purposes [9, 10, 11].

Regulatory and ethical frameworks

Clinical medical physics is regulated by various formal and professional structures. Since 1985, all medical radiological installations have been legally required to employ medical physics experts in accordance with the International Basic Safety Standards for Protection against Ionizing Radiation and for the Safety of Radiation Sources (IAEA Safety Series No. 9), the International Commission on Radiological Protection (ICRP), and the International Atomic Energy Agency (IAEA). Clinical medical physicists work with the International Organization for Standardization (ISO) and the International Electrotechnical Commission (IEC) to help develop standards for medical radiation.

Clinical medical physicists are also required to fulfill the criteria defined by the Joint Commission's Committee on the Accreditation of Healthcare Organizations (JCAHO) in order for hospitals and clinics to be accredited by the JCAHO. In the United States, the American Board of Radiology (ABR) defines the process for earning certification in therapeutic medical physics. The American Association of Physicists in Medicine (AAPM) has provided a mission statement, which can be summarized as follows: Clinical medical physicists are scientists trained in the application of physics to medicine, and they provide support for all aspects of medical imaging and radiation therapy. The AAPM also provides a recommended code of ethics for clinical medical physicists, which should guide these professionals as they work in a clinical setting.

The contribution to public safety and a society's overall health and well-being requires rigorous legal oversight, plus the formulation of accreditations, professional responsibilities, and codes of ethics. The compliance with all items on this checklist of whom to report and what laboratory systems to monitor ensures an evidence-based approach to the practice of clinical medical physics [12, 13, 14].

Chapter - 2

Atomic and Nuclear Physics for Medicine

Medical physics covers the applications of physical principles and techniques to the fields of medicine, biology, and healthcare. In the diagnostic realm, medical physicists are often responsible for ensuring the optimization of devices and equipment that use ionizing radiation, magnetic resonance, high- and low-frequency ultrasound, nuclear medicine, and other modalites. They collaborate with medical doctors, technicians, and other professionals in the interaction of these equipment with patients or test objects for the purpose of diagnosis and monitoring for checkpoint studies as well as with imaging equipment design and selection focused on dose and image quality. In the therapeutic realm, the medical physicist collaborates with medical doctors and other members supporting the use of techniques that employ ionising radiation or other agents for the purpose of treating cancer. The scope and competencies of medical physicists may also include dosage studies of radio-pharmaceuticals used in nuclear medicine, radio-biology and radiation protection, and quality-assurance testing of equipment and methods used for diagnosis and treatment.

Medical physics is a distinct field from general physics and engineering. Physics design and engineering are concerned fundamentally with the design of the equipment to specify how the equipment is constructed or how the equipment is operated. Medical physics, however, often does not influence the equipment designs or specifications but instead operates with what is provided and focuses on how to maximize the usage of what is provided. There are opportunities for dedicated devices or special modifications or attachments that are meant for either the diagnostic or therapeutic use, but the core specifications of the devices often remains unchanged and, therefore, the scope of activities are generally classed separately.

Medical physics is covered under several guiding documents that are required by laws in the respective countries. Medical physicists under cancer treatment activities are subjected to the Canadian Radiation Protection Regulations and the Canadian Nuclear Safety Commission (CNSC) are currently required to report any incident, medical or otherwise, where a dose is miscalculated or delivered incorrectly, even if the dose received does not impact patient safety. Medical physicists also works with other branches of physics such as nuclear, particle, x-ray, ultrasound, biophysics, chemical, and internal and external dosimetry. Medical Physicists Handbook of Canada of CMIU-54 Standard outlines medical physics specification across both diagnostic and therapy [15, 16, 17].

Atomic structure and electron interactions

Fundamental concepts of atomic structure serve as the foundation for the interactions of radiation with matter that define the processes in both diagnostics and therapeutic physics. Familiarity with the nucleus and its components is helpful for understanding modes of radioactive decay and the changes to the nucleus associated with radioactivity.

The nucleus of an atom consists of protons and neutrons (the nucleons). The atom is electrically neutral when the number of electrons equals the number of protons. The electrons orbit the nucleus at different distances and hence occupy different energy levels or shells. The projection of the orbital paths on the chosen coordinate plane shows the arrangement of those energy levels qualitatively; the energy levels are different at different distances from the nucleus. The electrons in an atom can be bound to the nucleus only if they possess discrete energy values. The energy required to remove an electron from a shell is inversely related to the probability of occupying that shell; therefore, in periodically stable atoms, the K shell is almost always the innermost and has the largest binding energy, while the outermost shell has the lowest binding energy. The energy levels of high atomic number (Z) atoms are relatively close; hence, Z and the energy of the interacting radiation determine the probabilities of interaction in the photoelectric effect [18, 19, 20].

Radioactive decay and nuclear transformations

Three fundamental decay modes are considered, namely β^- and β^+ decay and spontaneous fission. The properties of radioactive sources relevant for diagnostics and therapy are also described.

Radioactive decay

In diagnostic and therapeutic applications, radioactive nuclei decay according to Poisson statistics. Because types of radioactive decay are closely related to different types of nuclear transformation, it is convenient to use the following symbols:

- β^- decay:

$$\displaystyle _{Z}^{A}X \longrightarrow _{Z+1}^{A}Y + \overline{\nu} + e^{-} + \gamma$$

where $\overline{\nu}$ is the antineutrino. For β^- decay, the mass on the left-hand side is less than the mass on the right-hand side.

- β^+ decay:

$$\displaystyle _{Z}^{A}X \longrightarrow _{Z-1}^{A}Y + \nu + e^{+} + \gamma$$

where the neutron n is converted to a proton p. During the decay, a neutrino ν and positron e^{+} are emitted. For β^+ decay, the mass on the left-hand side is greater than the mass on the right-hand side.

- spontaneous fission:

$$\displaystyle {}_{Z}^{A}X \longrightarrow _{Z_1}^{A_1}Y_1 + _{Z_2}^{A_2}Y_2 + E.$$

where both the emitted fragments are in the region of stability. For spontaneous fission, the masses on both sides are equal.

The concept of activity describes the rate of decay of a radioactive source. The activity λ of a radioactive nuclide is defined as the number of spontaneous decays in unit time:

$$\displaystyle A(t) = -\frac{dN(t)}{dt} = \lambda N(t).$$

The activity is expressed in becquerels (Bq), where one Bq corresponds to one decay per second. The activity of a radioactive source is given by the product of the number of atoms remaining and the decay constant:

$$\displaystyle A(t) = \lambda N(t).$$

Radiopharmaceuticals and radionuclide generators

In nuclear medicine, the radiopharmaceuticals are very important as they are the carriers of radio isotopes. A radiopharmaceutical consists of a radioactive compound and a pharmaceutical component.

For radionuclide generators, the parent nuclide should be stable, easily available and feasible. For the radionuclide generator to be useful, the half-life of the parent nuclide should be much longer than that of the daughter nuclide during the radiopharmaceutical usage [21, 22, 23].

Types of radiation: α, β, γ, X-rays

Alpha (α), beta (β), gamma (γ) and X-ray radiation differ in the modes of production, the interactions taking place with matter and the ensuing clinical consequences. Alpha radiation is defined as the emission of a helium nucleus: two protons and two neutrons, the charges and masses that contribute to the very small penetration ability of alpha radiation. Particle production via a similar mechanism is not possible for beta radiation. Beta radiation consists of fast electrons or positrons emitted during radioactive decay, with a charge ±1, mass 1/1836 that of the proton and a penetration ability intermediate between gamma and alpha radiation, as the ionization density of its tracks and the predominant process of interaction with matter are both variations of situations for the simple particle. The ingredients necessary for the emission of gamma radiation from intensively excited nuclei undergoing de-excitation remain unaltered throughout those occurring during radioactive decay. The peculiar penetration and energy-loss properties of gamma radiation arise from the photon nature of the radiation. In X-ray production, rapid electrons suffer deceleration in metallic targets or change direction in collisions with atoms in the target material.

The different types of radiation interact with matter in different ways and at different energy ranges. These interactions depend strongly on the charge and mass of the particle as well as the energy of the impinging particle and relate thus to the ionization density of the radiation, to the probability and nature of the interactions with matter and to several clinical consequences. The comparison is illustrated in the characteristic ranges of the various types of radiation in tissues as well as in the interaction probabilities of the different types of radiation in matter [24, 25, 26, 27].

Radiation spectra and energy distributions

Are fundamental to understanding common sources used in medicine. The decay of radionuclides typically generates a spectrum of emitted energy values. Radiation sources can radiate low-energy gamma photons (e.g., ^241Am) that undergo multiple interactions before arriving at the detector, thus palpably shaping the measured distribution. The shape of the radiation spectrum in X-ray tubes also influences radiation detection and imaging quality. Efforts to consider these energy distributions culminate in the development of the Kerma concept, often viewed as a natural extension of the definition of absorbed dose.

The spectral characteristics of sources emitting X-ray and beta radiation, two types of radiation predominantly employed in medical applications, differ noticeably. Within the context of X-ray imaging, the following considerations become relevant: The X-ray spectrum is inherently wide; only some parts effectively contribute to imaging; the dose imparted to patients should be minimized, and using high-energy photons engenders greater patient dose associated with stronger penetrability into tissues and higher doses delivered to organs outside the region being diagnosed [28, 29, 30].

Chapter - 3

Radiation-Matter Interactions

In diagnostics, medical physicists are primarily involved in the selection and application of radiation protection methods, the reduction of patient and operator exposure to ionizing radiation, the choice and quality assurance of imaging systems and image processing algorithms, the optimization of nuclear medicine investigations, and the assessment of radiation dose and associated risk in diagnostic procedures. In radiation therapy, medical physicists play a key role in quality assurance and quality control, treatment planning and dose calculation, dose reporting and risk assessment, and the selection, commissioning and continuous quality assurance of radiation treatment devices. All medical physics activities are conducted according to established and publicly available protocols, preferably in an evidence-based manner and in accordance with recommendations from authoritative bodies such as the International Atomic Energy Agency, World Health Organization, American Association of Physicists in Medicine and Radiological Society of North America [31, 32, 33].

Photoelectric effect

The photoelectric effect results from the interaction of X-ray or gamma radiation with the electron shells of atoms and takes place predominantly in high-Z materials for low-energy radiation. The absorption of an incident photon results in the expulsion of a bound electron with a kinetic energy E_{KE} given by

$$E_{KE} = E_{\gamma} - E_B,$$

where E_γ is the incident photon energy and E_B is the electron binding energy. The probability for photoelectric absorption in matter per unit X-ray energy, $d\mu/\rho\ dE$, is given by

$$\frac{d\mu}{d\rho dE} \propto Z^S \frac{1}{{E}^{3}} \approx \aleph Z^5 \frac{1}{{E}^{3}},$$

for incident photon energies below approximately one-tenth of the atomic number Z. In diagnostic imaging systems, the photoelectric effect is important for the formation of bone and soft tissue contrast. Unlike scatter, which contributes to the background “noise” of an image, photoelectronic absorption generates net image information. However, because $d\mu/\rho\ dE$ increases with Z, the higher the atomic number of the absorbing material, the more probable its absorption of low-energy photons; this is the basis for the use of filters that preferentially remove low-energy photons.

The absorption of low-energy photons by the patient contributes substantially to radiation dose and the associated risk of deterministic tissue effects; the use of K-edge filters provided by the body for contrast-enhanced angiographic studies thus helps to reduce dose without significant loss of image quality. In therapy, the photoelectric effect, with its Z^3 dependence, is a major dose contributor in high-Z tissues near the surface where substantial doses are deposited with external beam therapy.

The photoelectric effect may also modify the radiation dose during brachytherapy in contact with high-Z applicators, especially at the skin-suture interface when iodine-125 sources are used [34].

Compton scattering

Is the interaction of an incident photon and an outer-shell electron. Both photon and electron energies after collision are given by the following relations, allowing one to assess the angles or energy shifts in Compton images or for dose estimation:

$$E'_\gamma=\frac{E_\gamma}{1+\frac{E_\gamma}{m_0c^2}(1 - \cos(\theta))}$$

$$E'_{e^-} = \frac{E_\gamma m_0c^2}{m_0c^2 + E_\gamma(1 - \cos(\theta))}$$

The Compton effect dominates the attenuation of clinical X-ray and γ-ray beams above about 20 keV, and its role generally increases with photon energy above this value. However, it is considered a nuisance effect in diagnostic radiology: it introduces radiation dose without contributing to image formation in projection techniques, lowers contrast by elevating noise levels, and reduces the accuracy of quantitative studies. Compton scattering is also responsible for reducing both the linear attenuation coefficient and the figure of merit in γ-cameras.

An important effect in the context of medical imaging arises because the scattered photon maintains only part of the incident energy. The lower energy of the scattered photon results in reduced attenuation for subsequent interactions and thus increases the probability of transmission through the patient without undergoing any further interaction. Consequently, high-energy scatter is the principal source of image degradation in projection imaging, forming the basis for scatter removal (generally through the application of a grid) without producing any corresponding improvement in patient image quality. In radiation therapy, Compton scatter invariably reduces the quality of dose calculations, necessitating the use of a correction factor (the ratio of the true scattered photons to those estimated by the primary-fluence method) [35, 36, 37].

Pair production

If a photon has at least 1.022 MeV of energy, it can reduce its energy by being transformed into an electron-positron pair, the two particles departing in approximately opposite directions (Chapter 11). This process, called pair production, occurs in the fields of nuclei, enables the photon to decelerate and is important in high-energy therapy and imaging. The energy required to create a particle-antiparticle pair is precisely 1.022 MeV, which is double the rest energy of an electron or positron. In general, pair production in the field of an atomic nucleus can occur when the energy of the photon information exceeds twice the rest energy.

The contribution of pair production to attenuation is significant not only because of the magnitude of the attenuation coefficients but also because it represents the only interaction of high-energy photons that produces matter. As a consequence, pair production is the only mechanism that leads to the creation of secondary charged particles, which can go on to initiate further ionizing interactions in the medium through which the high-energy radiation passes. In imaging applications, the conditioned avalanche of secondary charged particles caused by pair production plays a vital role in the detection process. In external-beam therapy, such an avalanche results in concentration of dose near the Bragg peak. In therapy with proton beams and, to a smaller extent, heavy-ion beams, a similar concentration of dose occurs because of proton energy loss via nuclear interactions [38, 39, 40].

Linear attenuation coefficients and HVL

The linear attenuation coefficient is a measure of the probability of photon interactions with matter, expressed as the fraction of incident intensity attenuated per unit length of matter. It varies according to photon energy and atomic number via different mechanisms, generally increasing with Z and photon energy in the diagnostic range due to dominance of the photoelectric effect, except for Z<12. For elements of similar atomic number, it usually increases with Z owing to the photoelectric effect's Z4 dependence. However, for pair production in high-energy photons, it depends mainly on Z. Linear attenuation coefficients are foundational for shielding calculations across the diagnostic and therapeutic photon energy range and are especially useful for radioprotection. In practice, half-value layers (HVLs), defined as the thickness of a defined material required to attenuate the beam to half its initial intensity, are more commonly quoted due to their direct influence on clinical operations.

The HVL is the most practical definition of photon attenuation, being the basis of clinical and administrative protocols for limiting patient and operator dose throughout medical practice. Its determination has become exceptionally swift in recent years, with dedicated software packages enabling direct input of unmonitored X-ray unit parameters for automatic HVL calculation from available tabulated data of known primary and scatter spectra [41, 42, 43, 44].

Chapter - 4

Radiation Dosimetry Principles

Medical physics has two principal branches: medical diagnostics and medical therapy. The domain of medical diagnostics exploits the interaction of energy—especially radiation—emmited, generated, or absorbed during physical or physio-pathological phenomena occurring in living systems, with a view to obtaining information useful for medical diagnosis. On the other hand, the field of medical therapy is concerned with the application of agents for the prevention, treatment, diagnosis, or cure of diseases affecting patients. Such information and therapy are very often precise.

Many areas of specialization have emerged from these two branches, including radiology, radiotherapy, nuclear medicine, and magnetic resonance imaging (MRI),among others. Nevertheless, a medical physicist is a professional with a recognized university degree who offers support to the other health professionals working in clinical or other practices, whereas for medical medical physics, the medical physicist is the individual recognized by law as competent in the field of medical radiation physics. These professionals are responsible for carrying out the quality assurance or quality control programmes of radiological or radiotherapy systems or nuclear medicine and MRI systems.

In the area of radiation protection, the role of the medical physicist is substantial because he is the professional who evaluates the factors that could originate deterministic or stochastic effects in patients, workers, and the general public. The objective is always to keep all exposures As Low As Reasonably Achievable (ALARA principle) [45, 46, 47].

Absorbed dose and dose equivalent

The absorbed dose quantifies the energy deposited by radiation in a volume of tissue and is represented by the quantity D given in units of Gray (Gy). One Gray corresponds to the absorption of one joule in one kilogram of tissue. Since the biological effects produced by equal absorbed doses at different radiation types may not be similar, the dose equivalent quantity H is defined by applying radiation weighting factors WR reflecting the relative biological effectiveness (RBE) of the various types of ionizing radiations.

Mathematically, the dose equivalent H (in sieverts, Sv) can be defined as:

$$H = D \cdot W_R$$

In medical physics practice, it is customary to separate the absorbed dose and radiation weighting factor in the definition of dose equivalent, so that D is expressed in Gy and WR in reciprocal units. When dose equivalent is defined for photons, and electrons, WR = 1, and the numerical values of H and D are equal.

In radiotherapy, the linear energy transfer values encountered in practical situations and the associated quality factors are usually known and it is customary to report only the absorbed dose in the treatment prescription. Modern planned treatment systems generate three-dimensional dose distributions from physical and biological data. Such distributions are expressed in absorbed dose units and are reported in Gy [48, 49, 50].

Kerma, LET, RBE, QF

The kerma concept quantifies the energy transferred from particulate radiation to charged particles per unit mass and is expressed in gray (Gy). In a medium exposed to fast neutrons and high-energy heavy particles, localized energy loss occurs. Depending on the secondary charged particle energy, tissue effects may not be evaluated using the absorbed dose due to low LET and relative biological effectiveness (RBE) considerations. The biological tissue effect is fully accounted for by the dose equivalent H, defined as the absorbed dose D multiplied by the quality factor QF, both values reported in gray. While QF accounts for tissue effects in neutron therapy, RBE expresses them in proton therapy. In proton therapy, QF varies with particle energy and tissue Z, while RBE approaches one with decreasing particle energy.

Clinical doses in diagnostic radiology are primarily defined in terms of kerma. Following Becker and Haines' radiobiological work, the recommendation for clinical reporting is to express doses in terms of absorbed dose (D), dose equivalent (H), or kerma (K), with the quality factor (QF) included for neutron beams. Kerma is usually reported in radiotherapy dose planning for generating isodose distributions and defining physical radiation doses. Neutron kerma in planning is applied with caution due to RBE in-vivo contouring and tumoral volume definition. The detailed absorbed dose calculation, when available, allows for tissue dose determination.

In clinical practice, dosimeters provide direct absorbed-dose estimation. Standard procedures and equipment calibration according to AAPM or IAEA protocols ensure dosimeter calibrated values represent absorbed dose. Absorbed-dose and kerma-related quantities can be expressed in conventional units (rad and rem). For radiation therapy, physical quantity definition and radiation dose measurement represent standard practice common to other electric technologies. Naturally evolved quality-assurance checks guarantee measurement accuracy with regard to patient and public safety [51, 52, 53, 54].

Dosimeters: ionization chambers, TLDs, OSLDs

Dosimetry is vital to ensuring that patients receive an appropriate amount of radiation, whether the application is diagnosis or therapeutics. Over- or under-dosing can have serious consequences, so it is crucial that equipment and techniques be tested regularly and that any necessary adjustments are made. To ensure that the equipment used in clinical practice delivers the intended dose, its output must be routinely tested. In addition, radiation therapy treatment plans are typically verified using *in vivo* dosimetry techniques. To help guarantee that these processes are performed accurately, dosimeters are used. The most common types of dosimeters are ionization chambers, thermoluminescent dosimeters, and optically stimulated luminescent dosimeters.

Ionization chambers are commonly used to measure absorbed dose and air kerma in the medical context, particularly in radiology and radiation therapy. These physics quantities can also be measured using thermoluminescent dosimeters or optically stimulated luminescent dosimeters. TLDs and OSLDs are also used for *in vivo* dosimetry in external beam radiation therapy. Ionization chambers are mainly used in absolute dosimetry for calibration of secondary and primary standards or in the evaluation of clinical dosimetry protocols. TLDs and OSLDs, on the other hand, are typically used in relative dosimetry for clinical quality assurance and quality control, including the use of laboratory-checking dosimeters [55, 56, 57].

Calibration standards and protocols

Standardized calibration protocols ensure accurate absorbed dose measurement in radiology and radiotherapy. In diagnostic radiology, radiation output from X-ray systems is usually calibrated as air kerma in a soft tissue approximation. In radiotherapy, clinical linear accelerator beam output is established using water as the reference medium for a primary standard calorimeter. The International System for Units (SI) and International Committee for Weights and Measures (BIPM) recommends primary calibration of radiation dosimetry equipment in gray (Gy) for absorbed dose (AD) and gray per joule (Gy/J) for energy absorbed in matter. The appropriate transfer for air kerma calibration is gray in kilograms per second squared (Kg/s2). The International Atomic Energy Agency (IAEA) conducts primary calibration of absorbed dose in water, with a secondary standard series of ionisation chambers.

In both diagnostic radiology and radiotherapy, the response of clinical ionisation chambers must be cross-calibrated at the radiation qualities normally encountered during clinical practice. For radiological applications, this is performed as a kerma calibration factor, ka, in gray per air kerma (Gy/Kg/s2) at a traceable national or regional metrology institute for radiation physics (NMI). Using the chamber calibration factor, air kerma absorbed dose can be determined from the measured response of the chamber: (D−kam)

For radiotherapy dosimetry, AAPM-recommended protocols are the primary methods for beam output of shielding calculations, or if used for reference dosimetry, must use a protocol appropriate for the selected dosimeter type.

Chapter - 5

X-Ray Production and Properties

Medical physics is subdivided into diagnostic and therapeutic branches reflecting the essential purposes of radiation in medicine. In diagnostic imaging, medical physics examines the generation, behaviour and detection of ionising and non-ionising radiation. In addition to ensuring the safety and correct operation of these systems, medical physics investigates how imaging quality may be improved and how exposure of the examined patient can be optimised.

In therapeutic radiological physics, the word ‘radiation’ in the wider sense refers to any technique whereby energy of any type forms the basis of treatment. Thus the governing elements of therapeutic physics involve the generation of the energy and the delivery of that energy into diseased tissues whilst attempting to leave normal tissue as undamaged as possible [58, 59, 60].

Bremsstrahlung and characteristic radiation

In any X-ray tube, the relatively high speed kinetic energy of the incident electrons is converted to electromagnetic radiation at decelerating points on the electron cloud of the anode atomic nuclei, producing Bremsstrahlung widely spread over a wide spectrum of energies. The high-energy part of this bremsstrahlung radiation is responsible for most of the dose absorption in diagnostic radiology, as well as for its production and tissue effect in radiation therapy. Characteristic radiation, on the other hand, is emitted as narrow peaks associated with the different ionization transitions of electrons falling from higher energy levels into the vacancy created by the impinging electron. These peaks are much more intense than the corresponding areas of the adjacent bremsstrahlung spectrum.

Thorium, Se, U, W, and Pt are widely used materials. W is the favored material because it has a suitable melting point, is sufficiently abundant, and allows, through the presence of other chemical elements, the modification of the characteristic X-rays; it also absorbs the majority of the highly penetrating bremsstrahlung part of the spectrum without significantly attenuating the lower energies. These, on the other hand, have a low conversion efficiency; for a cold cathode the efficiency of charged particle to bremsstrahlung energy conversion can be as low as 5%, whereas a cool tube emits proportionately more of the lower energies with a conversion efficiency of around 30%. The absorbed dose is given by $D \cdot \int E\ F(E)\ dE\ \sigma B(E)$, where $F(E)$ is the bremsstrahlung spectrum. The ineffective sensitivity of the skin to low-energy X-rays has led to the use of cooling systems at the anode under superimposed centrifugal forces, thereby making use of the very high specific heat of liquids [61, 62, 63].

X-ray tube components

An X-ray tube produces penetrating radiation by bombarding a target with high-speed electrons. The X-ray tube is designed to facilitate electron generation and acceleration, direct the electron beam onto the target, and contain components that assure operator and patient safety. An X-ray tube contains two principal electrodes: the cathode, which provides the electron source, and the anode, which allows photons to escape while dissipating heat generated by the incoming electron beam. The inner parts of the tube that house the anode and cathode are evacuated.

X-rays are produced primarily by the deceleration (bremsstrahlung) of electrons passing in close proximity to nuclei. Characteristic radiation is generated when high-speed electrons collide with orbital electrons in the target material and cause them to be ejected from the atom. The resulting vacancy is then filled by a higher-level shell electron emitting a photon. X-ray-tube design maximizes photon production per acceleration voltage and minimizes the number of electrons striking the anode that do not contribute to radiation production [64, 65, 66].

Spectral shaping and filtration

Influence patient dose and image quality by modifying emitted x-ray photon energies. Filtration refers to interposing material in the x-ray path (typically Al), while spectral shaping encompasses aperture and inherent filtration effects. Although tube anode materials absorb low-energy incident photons, those with sufficiently high energy subsequently generate characteristic radiation of similar energy. Thus, increasing the atomic number raises the characteristic photon fraction and average tube energy; albeit advantageous because of increased beam penetration, this effect leads to an escalation of patient dose. For these reasons, molybdenum or rhodium anodes are used in mammography, where lower energy levels are acceptable.

Grids are employed to mitigate scatter effects in projection radiography. Within the primary beam, x-ray scattering (i.e., Compton interactions) increases with patient thickness; scatter contributes to image fog and, at sufficiently high levels, to exposure safety concerns. Consequently, examination of thicker anatomical areas requires radiographic grids to lessen scatter fraction incident upon the detector. Grids consist of alternating radiopaque and radio-transparent materials capable of absorbing scattered but not primary photons. Grid geometry determines grid ratio—defined as the ratio of the height of the radiopaque layer to its separation from the adjacent layer—and therefore its selectivity. Indeed, increased grid ratio shifts the grid's transmission profile toward higher photon energies [67, 68, 69].

Tube rating charts and safety limits

A tube rating chart indicates the safe operating conditions of an X-ray tube for given exposure times. By defining electrical parameters (incoming voltage, tube current), filtration level, and anatomical focus (from a specified source), these charts help assess the extent of tube exposure and prevent damage during clinical examinations. The maximum heat on the anode and the degree of permissible risk are considered.

An X-ray tube resembles a thermometer: increasing heating leads to thermal expansion and melting. When demand exceeds production, damage occurs. Tube rating charts and time-temperature graphs integrate information from manufacturers and equipment developers into toolkits for maximum tube operation and clinical protocols.

Plotting allowable parameters against gas-flow temperature enables rapid grade assessments.

Calculations based on an electrostatic field show that the production of X-rays is associated with the dissipation of enormous amounts of energy. In a diagnostic tube, only ~0.5% of the electrical energy is used for X-ray production. Only 1-5% of the beam consists of radiation with sufficient energy to penetrate the patient with decent transmission and be detected. Consequently, considerable energy must be dissipated in the anode during even a short examination [70, 71, 72].

Chapter - 6

Radiographic Imaging Physics

Medical physics is a broad multidisciplinary field, which supports in maintaining safety standards in diagnostic and therapeutic procedures using ionising radiation. In diagnostics, various imaging techniques such as radiography, fluoroscopy, CT, MRI, nuclear medicine and ultrasound are employed for disease detection. The emphasis in medical screening physics is to ensure safety, importance being given to minimisation of radiation dose to patient and to the personnel involved, not affecting the quality of the diagnostic information. For therapy, modalities using ionising radiation, like external beam irradiation, brachytherapy and neutron therapy are used. Medical therapy physics concentrates on dose delivery to tumour while minimising the dosage delivered to neighbouring healthy tissues.

The roles of physicists in these areas differ because of technical emphasis and procedures peculiar to the different branches. The medical screening physicist lays down required information for workings in each application. He sets limits for controlling images, which may grant low dose for the same information obtained in a poorer image. The approval of any new device for clinical trials is at his investigation as the assessment is more of an engineering one but with radiation safety as a constant wish. The clinical physicist, apart from advising on the devices of use, is also in charge of the normal operation, super-vision of quality control checks and suggesting any better approaches or make shifts to improve the working [73, 74, 75, 76].

Image formation in projection radiography

Image formation in the projection radiography process, in which two-dimensional information is obtained from three-dimensional objects, is based on the attenuation of X-rays in the patient. Geometric properties of the system are responsible for magnification. The size and relative position of the X-ray source and the detector surface determine the degree of oversizing of the projected image. The spatial resolution of the system is defined by the size of the unsharpness seen in image proximity to the X-ray source. Use of a grid to suppress scatter radiation improves image contrast but also adds patient dose. Other scatter reduction techniques may be employed to optimize image quality.

The design of an X-ray imaging system defines the image formation process, the uncertainty of the spatial resolution, and the magnitude of the scattered radiation relative to the useful signal. The first two aspects have been discussed, while the impact of the scattered radiation on the quality of the resulting image has yet to be addressed. The Use of a grid, a device positioned between the patient and the X-ray detector, permits a considerable reduction of scatter radiation at the expense of an increase of the useful beam dose to the patient. Scatter can also be reduced through the use of thin, low-Z materials (such as air) in the vicinity of the patient and the use of oblique angled X-ray beams [77, 78, 79].

Geometric factors and magnification

Image formation processes in projection radiography differ fundamentally from those in imaging modalities based on spatial resolution, such as CT or MRI. A simple pinhole view could be obtained by projecting the X-ray source conditions through the object on to a film sensitive only to incoming radiation. In such a system, the object appears smaller than its original size and features that are sufficiently small in size may become invisible. However, the magnification is inversely proportional to the distance of the object from the film. Therefore, an increase in the closeness of the object to film results in a larger indicated size and a higher chance of the image becoming susceptible to partial volume effects.

Consider a more practical situation in which a point source, S, at a given distance from the film, F, projects the object, O, in a direction, R, perpendicular to its surfaces. If, as is normally the case, the object is placed between the source and the film, the induced pinhole image, O’, will be larger than the real object and not free from the effects of geometrical errors. These errors, expressed in terms of magnification factors, can be quantified independently of those arising from the size of the object using the distances: Source to Object (d_1), Object to Film (d_2) and Source to Film (d_3). In this case, the magnifying factor, M, can be expressed as:

$$M = (O'F)/(OF) = d_3/d_1$$

where O'F and OF are the distances of the pinhole image and the object, respectively, from the film along the direction of the pinhole.

It is customary in projection radiography to view the process as a "quantum casting" of the object in which the intensity of film exposure at any point remains proportional to the "density" of quanta emitted from the corresponding point of the object in the direction of transmission. The film represents a map of the absorption distribution in the object projected along rays from the source arranged so that, within limits, the distortions due to the geometry of the pinhole projection process are negligible [80, 81, 82, 83].

Grids and scatter reduction techniques

Projection radiography (X-ray), fluoroscopy, and cone-beam computed tomography are affected by the Compton scattering of incident X-ray photons in the patient. Scatter reduces radiographic contrast, and measures of image quality can be impaired to an unacceptable extent in normal clinical use. Grids are employed to reduce the proportion of scatter reaching the detector, improving image quality. These devices use a combination of absorption and transmission to reduce scatter while allowing primary radiation to pass, although this process can increase patient dose. Other scatter reduction techniques are also employed to mitigate this effect.

In X-ray systems, Compton scattering is the primary source of image degradation due to its prevalence among other interaction mechanisms. The dominant directionality of scatter production within the body results in a lack of shadowing and therefore reduced contrast. The use of grids is a common approach to improving contrast by increasing the relative magnitude of transmitted versus scattered radiation [84, 85, 86].

Image quality metrics (resolution, contrast, noise)

In image formation, primary interest lies in the distinction among various tissues. The visibility of such differences depends on three fundamental image quality metrics: spatial resolution, contrast, and noise. All three are directly related to the underlying physical processes. The following sections provide qualitative and quantitative definitions of each metric and discuss their clinical significance.

Spatial resolution designates the level of detail in the spatial distribution of the detected signal. In projection radiography, the ultimate resolution depends on the size of the focal spot, the distance to the image receptor, and the magnification factor. Motion of the patient or the imaging system relative to the image receptor produces unsharpness that limits the resolution in a space-variant manner and is quantified by the modulation transfer function of the system. Image contrast quantifies the visibility of an object’s edges and is defined as the difference in signal value between adjacent regions divided by the average of those values. Contrast is directly determined by the underlying physical processes, such as the photoelectric effect in X-ray imaging. Systems that operate in the low-energy region (high-Z absorber, low photon energy) exhibit high contrast but high radiation dose; systems that operate far into the Compton scattering region can exchange contrast for dose by selecting a higher photon energy. Finally, noise represents the random fluctuations in the signal. The quantification of noise can be achieved by measuring the variance of the signal in a uniform area of the image. As with contrast, noise levels are often inversely correlated with radiation dose [87, 88, 89].

Chapter - 7

Fluoroscopy and Mobile X-Ray Systems

Medical physics has two main branches, Therapeu-tic and Diagnostic, based on its application in the field of medicine. Therapeutic physics includes the application of various physical principles in understanding and developing techniques/ methods for the safe and effective treatment of diseases and disorders. The knowledge and expertise in robotics, automation, vacuum technology, computational modeling, etc. are playing a vital role in developing new Advanced Radiotherapy Machines & Treatment planning Systems.

Diagnosis of diseases and disorders has always remained a challenge for the medical professionals. Though they can get information regarding the body system from the laboratory tests, necessary investigations are suggested to get correct information about the internal body structures. The role of therapeutic physicists is significant in preparing these machines for diagnosis. The Diagnostic Medical Physicists have the responsibility of ensuring the quality of medical imaging equipment that are used for diagnostic purposes. Medical Imaging equipment such as CT Scan, MRI, Ultrasound, Nuclear Scans, etc. pose certain dangers to the patients. Medical Physicists work closely with Clinicians to reduce the dangers and to make the diagnosis safe and sound [90, 91, 92].

Image intensifiers vs. flat-panel detectors

Image intensifiers and flat-panel detectors are the two most ubiquitous technologies used in radiography and fluoroscopy. Both systems have undergone extensive development over the years, guided by advances in imaging detector technology and underlying optical principles. Image intensifiers remain the technology of choice for general-purpose fluoroscopic applications, providing high-output images across a wide dynamic range. Flat-panel detectors offer increased versatility and compactness, facilitating the development of interventional and hybrid systems. However, optimization of image quality can still prove challenging, particularly for mobile and portable designs where the constraints on physical size compromise optimal performance.

The introduction of pulse fluoroscopic systems, which minimize the exposure rate during periods of non-visualization, has reduced patient doses in pediatric examinations but introduces challenges in the assessment of vascular flow. Likewise, the increasing use of C-arm systems has created a requirement for technological design that considers not only imaging quality but also the effects of varying geometry on image artifacts and the patient dose delivered [93, 94, 95].

Pulsed fluoroscopy and dose reduction

In pulsed fluoroscopy, the X-ray beam is activated intermittently using an alternating signal to the tube, producing a sequence of short exposure bursts. Together with a digital/flat-panel matrix configured for high switching speed, this technique significantly reduces patient dose compared with continuous operation. The reduced continuous dose rate allows the application of a variable low-speed mode of quiescence for the fluoroscopy operator, often in conjunction with a shutter or grid-controlled image intensifier. Main disadvantages of pulsed fluoroscopy are related to the inevitable increase in noise and reduction of analysis fidelity in dynamic sequences. These aspects can be partially alleviated by adopting lower frame rates.

In addition to pulsed exposures, other dose reduction strategies exist. The sensitivity of the image detector to radiation can be modified physically using an image intensifier, increasing the SNR of the radiographic process without changing the dose. For fluoroscopic procedures, permanent BBO crystals or a ceiling-mounted image intensifier can be used; in addition to lowering (or allowing for more visual fidelity at) higher fluoroscopy currents, these solutions may also reduce the need for continuous tube dosimetry. These advantages must be weighed against the costs and installation requirements of these devices. Similar considerations apply to the use of asynchronous C-arm fluoroscopy systems [96, 97, 98].

C-arm physics

The design of C-arms facilitates three-dimensional motion (translation and rotational) of both the imaging system and the patient. This flexibility may produce and large images of the examined anatomy with a geometry suitable for radiography and fluoroscopy. When fluorography is operated in continuous mode, the large quantity of scattered radiation produced by this high-radiation-quantity procedure poses a significant risk for both patient and staff. The X-ray source has to be placed as close to the image intensifier as possible in order to confine the volume irradiated in the patient and thus reduce the amount of scatter radiation produced. Different techniques for adequate scatter reduction can be applied: continuous suboptimal filtration of the beam, installation of a suitable collimator, use of dual-focus beams, operation of the fluoroscopic system in very short puffs (pulsed fluoroscopy). The significant reduction of obscured biological information achieved by continuous C-arm fluorography for the thoracic region, combined with long fluorographic recordings for the rest of the body, justifies the trade-off in dose. The nadir position of the C-arm does not allow the use of girder guidance, and this could produce excessive patient dose when the C-arm is not used with care or with the help of an experienced operator. Therefore it is absolutely necessary to install motion warning devices. The use of a large, preferably hemispherical image intensifier removes the need for large correction of distortion, provided the system is actively used in the direction of the antero-posterior portion of the body. Automatic optical correction of image distortion for both film recording and viewing is very practical.

The rolling-bearing-guided lateral C-arm features different advantages and drawbacks compared to the girder-guided version. By not requiring an internal canopy, this configuration is better suited for patients whose treatment must be performed in prolonged sitting position, while its reduced height when compared to a girder reduces the radiation scattering from patient to operating team. On the other hand, the use of a small image intensifier in the vertical fluorography of the abdomen increases the irradiated volume of the patient, and therefore the risk of biological damage from radiation. A comparison of the two systems applied to the thorax confirms the advantage of the girder-guided system, despite a large family of biological corrections for scatter radiation. In any case, the magnitude of the risk warranting continuous convergence towards the surface of the body assures useful biological information on the underlying area [99, 100, 101].

Clinical applications

In tank-based CT systems, the X-ray tube and the detector rotate around the patient to cover a circular or elliptical path. Data acquisition involves multiple consecutive rotations, generating a helical structure of scanned data with varying levels of angular data density. CT detectors are typically scintillation detectors, utilizing a scintillation material with short decay time and subsequent optical fibers and photon detectors. Reconstruction algorithms, relying on principles of linear systems and considering pixel intensities as projections, are applied in two-dimensional Fourier transform spaces or with explicit distance correction. Filter back-projection and iterative reconstruction are common techniques in modern multislice CT.

To support patient safety, specifically dose monitoring and optimization, CT dose indices are defined and measured. The CTDI estimates the absorbed dose for a continuous scan of an infinite phantom, while the DLP provides a dose-weighted length along the scanning axis. The Kerma area product, sensitive to the degree of collimation, serves a similar role in fluoroscopy dose monitoring.

A principle of image formation in projection radiography is that the intrinsic properties of the two-dimensional object, namely the topology and attenuation, dictate the formation of an image or shadow, while the distances between the source, object, and detector determine the absolute scale of the image. The distinct nature of the image pattern generated by the structure is described by a simple physical or geometric law [102, 103, 104].

Chapter - 8

Computed Tomography (CT) Physics

Medical Physics consists of two branches: the first relates to the physics applied to diagnostic medical imaging and support in diagnosis; the second to the radiological therapy and the physics associated to the therapeutic uses of ionizing radiation.

Medical Physics applied to diagnostic techniques includes aspects related to the production and interaction of diagnostic imaging. It also includes the formation and generation of different types of physical signal in the terms of diagnostic quality in relation to dose, the design and development of detection systems in the various modalities and the analysis of given information.

The Therapeutic Physics deal with research and applications of ionizing radiation for treatment purposes and the proposed physical processes applied in the intervention or restoration of malfunction in organs or in the entire organism. The different therapeutic methods use the radiation emitted and the interaction of these with matter. External radiation therapy makes use of very high energy beams from linear accelerators; brachytherapy introduces radiopharmaceuticals from the surface to the deepest regions of the organism; internal radiotherapy uses radiation of long half-life for the treatment of different diseases.

In the area of Medical Physics there are also the assistants professionals that guarantee the proper working of the equipment, assuring the quality of the diagnostic images and calibrating all devices used for verifying the doses in radiotherapy, for radiological protection, for quality control and for dosimetry in nuclear medicine.

Spiral and multislice CT technology

Modern imaging instruments enable the collection of multidimensional data. In spiral CT systems, a continuously rotating

X-ray tube and a linear array of detectors move along the z-axis during the scan. Multislice systems fully exploit this principle, using multiple detector rows for simultaneous data collection at several planes. They are faster than single-slice spiral scanners and allow thin-slice acquisition for improved surface or volumetric resolution.

Data acquisition occurs as a function of the angular coordinate. Sorted along a single axis, with each view corresponding to a radial coordinate defined by the compass angle (i.e. the angle of the data relative to the horizontal axis), the data are used to reconstruct the final image. Further processing may yield multiplanar reformations or virtual endoscopic views.

Scintillation detectors, solid-state detectors and xenon gas chamber detectors have been employed. Data acquisition is most commonly achieved by a fan beam, but cone-beam designs exist. Detector design tending towards greater efficiency and speed has allowed significant reduction in radiation dose, especially with the use of wide-area, low-GRE cone-beam acquisition techniques.temperature changes.

CT detectors and data acquisition

Multislice and spiral CT technology allows three-dimensional imaging and has applications beyond axial radiography, such as fluoroscopy, angiography, and cardiac imaging. An understanding of multislice configurations is essential for optimizing dose in these diverse applications.

In multislice systems, scintillator-based detector elements (pixels) sample the X-ray data in parallel, whereas in a spiral system, individual detector elements (singleslice systems) are not used in parallel. The multislice configuration therefore facilitates a wide cone-beam geometry, allowing all view angles in a cone of given size to be acquired in a small fraction of a rotation. The fundamental principle of multislice and spiral CT data acquisition is that additional rows of detectors accelerate data acquisition and allow the use of wider beams.

Reconstruction algorithms

In computed tomography (CT), the measured projection data constitute an incomplete set of the two-dimensional Fourier transform coefficients of the object's attenuation distribution. Consequently, the reconstruction problem is often ill-posed, and artefacts may arise even under ideal conditions. Different mathematical formulations underlying the common reconstruction algorithms are summarised. A reconstruction algorithm for spiral CT differs from that used in conventional CT mainly in the treatment of the projection data. Two-dimensional Fourier transformation of a single rotation set of projection data results in a dual array (the projection data were sampled at nonuniform intervals along the first coordinate). The performances of different algorithms in restoring the original image from simulated projection data of a numerical teapot phantom were compared.

CT reconstruction algorithms can be grouped into analytical methods (typically implemented in commercial scanners) and iterative methods (having qualified to be included in commercially available packages). An analytical reconstruction algorithm faithfully restores the attenuation distribution from an infinite number of noiseless projection data. With real data, the performance is influenced by noise, incomplete data, and mathematical approximations such as the use of a finite source size, non-circular scanning path, and non-parallel projection geometry. An iterative reconstruction algorithm is an operator-based method. An image of the object is simulated and projected; the simulated projection data are compared with the measured data, and corrections are made to the simulated image. This process is repeated until the simulated projection data match the measured data of the scintillator.

CT dose indices (CTDI, DLP)

Two key indices, CTDI and DLP, are used for radiation dose monitoring and optimization. Their definitions and clinical significance are discussed below, together with current international recommendations and guidelines.

CTDI is widely employed for characterizing patient dose in a CT examination. The principle behind CTDI involves measuring the air dose produced during a series of irradiations from a simple source. If the CT scan length is L and the radiation dose measured with a 100-mm pencil ionization chamber is denoted by TD (i.e. the average dose over that length), CTDI can be defined as CTDI = TD, a single value that does not require considering irradiation length. Measurement of the dose distribution at the level of the pencil chamber gives the variation of dose over the CTDI volume, which is always a cylinder of diameter and the same length as that of the sensitive volume of the pencil chamber. The relation between CTDI and the average dose in the central region of the phantom can then be determined. The AAPM Report No. 96 provides the following clinical recommendation for CTDI.

DLP is defined as the integral of CTDI over the length of the scan. It is the product of CTDI and scan length, and it is expressed in mGy cm. An estimation of DLP for a thoracic CT examination is between 300 and 1000 mGy cm. When radiologists discuss radiation dose in CT, they normally refer to DLP.

Chapter - 9

Ultrasound Imaging Physics

Medical physics encompasses the application of physics principles and methods in medicine but is further divided into two main areas of specialization: diagnostic imaging and radiation therapy.

Diagnostic imaging medical physics supports all methods of imaging to detect diseases and pathologies using non-ionizing, ionizing, and radioactive radiation, and covers X-ray-based imaging (general radiography, computed tomography), nuclear medicine (PLAN gamma camera, SPECT, PET), ultrasound, and magnetic resonance. The emphasis is on image quality, patient and personnel safety, and operations with the lowest possible radiation dose, without compromising image quality. The medical physics committee provides information on optimization and quality assurance and control programs for diagnostic imaging equipment. Radiation therapy medical physics focuses on the application of radiation for treating tumors, using ionizing radiation (photons, electrons) and radioisotopes for brachytherapy. Medical physicists ensure safety and quality of treatment techniques and computerized treatment planning systems, and the accuracy of treatment delivery to a patient. They are essential for the commissioning and quality assurance of complex radiotherapy machines and treatment planning systems.

Medical physics in radiation therapy integrates principles of radiation genetics and radiobiology, radiophysics, radiation protection, and clinical imaging, including techniques for kV and MV imaging, cone-beam computed tomography, and surface imaging systems, and contributes to all aspects of patient safety. Quality assurance of complex radiation therapy machines, implantable radioactive sources, computerized treatment planning systems, and *in vivo* dosimetry for high-dose-rate brachytherapy are fundamental to safe practice in radiotherapy [105, 106, 107].

Acoustic waves and propagation

Sound is a mechanical longitudinal wave, the propagation of compressions and rarefactions in an elastic medium. When modulation occurs in the time or frequency of a sound wave, it is referred to as an acoustic signal. When the source vibrates at frequencies in the range of 20 Hz to 20 kHz, the sound produced is audible to humans. The sound above 20 kHz is called ultrasound, and the sound below 20 Hz is known as infrasound.

Ultrasound is produced by the construction of high-frequency sound waves through repeated mechanical vibrations. These vibrations are sensed through a piezoelectric crystal, which also acts as a transducer. In this process, when the pressure is applied to the crystal, a mechanical wave is generated and also converted back into an electrical signal. The frequency of an ultrasound wave is typically 10-1,000 MHz. These mechanical waves use the elastical property of any medium for their propagation. The speed of sound depends on the density of the medium; for example, it is approximately 344 m/s in air, 1,535 m/s in water, and 5,169 m/s in steel. The speed of sound varies inversely with the density of the medium [108, 109, 110].

Transducers and beam formation

Acoustic energy is generated through mechanical pressure variations applied to the material of piezoelectric crystals; during wave propagation, the energy density is related to density, sound velocity, and particle velocity. The specific acoustic impedance determines the amount of freely transmitted energy through a fluid-tissue interface, indicating that reflections occur at structures with a mismatch of acoustic impedance. Transducer elements are piezoelectric crystals activated by an oscillator signal, while the transmitter-receiver functions are performed by separate elements or different phase operation of the same elements, generating a tapering beam with various widths. Each emitted ultrasonic wave is usable in return; the time-lapse Δt is registered in the receiving circuit. Volume is scanned by subsequently activating transducers along a straight line with known intervals.

A Doppler frequency shift indicates blood flow within the vessel.

A small device is positioned on the arteries; ultrasound p waves excite the tissues; blood cells scatter the waves, which return at slightly different frequencies. The apparent move of the Doppler sensor reflects the blood cell movements. Negative results mean no flow. It provides information about flow speed; flow direction is indicated by colors. Vessel occlusions cause typical color flows, while regrowth induces more complex directional patterns. Artifacts are unintended ultrasound signals arising from interference. Limitations on angles are essential; a narrow range provides the best information. In contrast, wide angles provide complementary but less precise results; however, they still affect diagnostics. Simple techniques with colors often correct these effects [111, 112, 113].

Doppler techniques

Allow the detection of motion-induced frequency shifts and are utilized in several clinical applications. A transducer operating in receive-only mode is capable of detecting flows toward or away from the transducer. The simplest application uses a single transducer to measure flow rates in vessels along a parallel direction. The actual direction of blood flow is detected by operating the transducer in duplex mode: a second transducer is dedicated to imaging the vessels, which can also be located at a perpendicular angle with a dedicated probe.

Since Doppler techniques exploit the departure of detected frequency from the emitted one, the presence of flock artifacts can corrupt the interpretation. Modifications in the emitted pulse can mitigate the problem; nevertheless, these artifacts are hard to correct in 2D modes, which use a single transducer to emit and receive the pulse. The incorporation of diverging beams and a speed threshold (often close to zero) can further reduce misunderstandings [114, 115, 116].

Image artifacts and correction

Despite its numerous advantages, ultrasound imaging is affected by a variety of artifacts that can obscure or complicate image interpretation. Artifacts can originate from the physical limitations of ultrasound imaging or from the practical consequences of the transducer's position relative to the examined structure or tissue.

Although some artifacts can be useful for the correct interpretation of the image, others may obscure anatomical structures or mimic pathology, which, in tests with ultrasound contrast agents, are known as “mistakes.” Numerous studies have characterized these artifacts; the most important ones are described here.

Acoustic shadowing occurs when a highly reflecting surface or a strongly attenuating medium lies in the path of the beam. The significant loss of intensity in the distal area produces a blackened region. This artifact is commonly seen behind bones, gallstones, kidney stones, or air-filled lungs. Acoustic shadowing occurs excessively in the pan-artery when a large amount of atherosclerotic plaque is present. It can also make it difficult to detect the true depth of a structure, as in color-Doppler imaging, for example, the detection of low-velocity flow behind plaque caused by strong reflection is challenging.

In medical ultrasound, a significant drawback of traditional beam formation techniques is sidelobe energy, which reduces imaging contrast and resolution. Sidelobe energy can be a problem in the detection of low-contrast objects. The presence of evasive decoherence points can transform sidelobe energy distribution into major lobe energy distribution, which, under certain specific conditions, have a simple theoretical interpretation. The response of any imaging modality has the challenging problem of measuring and accounting for decoherence (strong sidelobe energy) in its treatment, detection, or classification. Various methods introduced to solve this problem at the practice level emphasize the control of decoherence-point-introducing perturbations, errors, or effects during the measurement and may incorporate decoherence metrics into their detection functions using machine learning.

Modern ultrasound methods allow a large range of parameters to be set. Dynamic alteration of parameters according to scene statistics is thus possible, enabling energy to be concentrated where needed and reducing mistakes elsewhere. Scene complexity can be judged by means of weighted local classifications applied, for example, to detection, and multiclassifiers can be applied dynamically using, for instance, energy estimates [117, 118, 119].

Chapter - 10

Magnetic Resonance Imaging (MRI) Fundamentals

Two primary branches of medical physics exist: diagnostic (or investigative) medical physics and therapeutic (or treatment) medical physics. The primary role of a diagnostic medical physicist is to ensure the provision of medical images of the required diagnostic quality for patients undergoing X-ray or magnetic resonance imaging examinations. The role in diagnostic nuclear medicine includes ensuring that the images acquired are of sufficient quality to allow an unambiguous diagnosis. The role in a diagnostic scientific metabolic laboratory (a nuclear medicine laboratory capable of metabolically investigating human tissue and blood products) is to ensure accurate measurement of the radioactivity of human samples.

The role of a therapeutic medical physicist is to ensure that therapeutic radiological or radiation therapy treatments provide a clinical outcome that is optimally beneficial to the patient, while endeavouring to ensure that the risk of radiation injury is acceptably low. The implementation of such treatments requires, among other things, that the radiation dose distributions delivered to patients are as intended, that the administration of a brachytherapy source is performed safely, that the output of a radiation therapy unit is precisely known, that the radiation dose absorbed by the patient is monitored during the course of treatment, and that established quality control protocols are performed [120, 9, 1, 121].

Nuclear magnetization and precession

Under the influence of a strong static magnetic field B0, the magnetic moments associated with atomic nuclei in a biological sample tend to align with the direction of the field, giving rise to a net nuclear magnetization M. Once the system has reached equilibrium, the net magnetization points along B0. A brief radio-frequency (RF) pulse applied orthogonally to B0 tips the magnetization away from equilibrium, generating a non-equilibrium signal that can be detected and imaged. After the tip, the longitudinal component of the magnetization recovers exponentially, characterized by the T1 relaxation time; the transverse component decays exponentially, characterized by the T2 or T2* relaxation.

T1 and T2 times depend on tissue physical properties as well as chemical composition. For example, differences between fat and water T1 values are exploited to improve fat suppression in MRI. The magnitude and temporal decay of the MR signal following a pulse depends on the density of magnetic moments in the sample and the angle of the perturbation produced by the RF pulse.

In twisted-spine MRI, a phase-preserving radiofrequency pulse is used to create a signal for every region containing fat, which decays rapidly and serves to mask neighbouring signals that decay relatively slowly. Because precession frequency is also determined by the strength of B0, acquisition techniques can be designed to generate image contrast based on differences in precession frequency, providing the T2 contrast that is used in twisted-spine imaging [122, 123, 124].

Relaxation times (T_1, T_2, T_2)

Magnetic resonance imaging employs pulsed radiofrequency signals to detect nuclear magnetization within the tissue and to induce the precession of the nuclear spins. The temporal evolution of the nuclear magnetization can be described in terms of two relaxation times, T1 and T2, which characterize the recovery of the longitudinal and transverse magnetization components, respectively. A third relaxation time, T2*, is also defined, describing the true decay time of the transverse nuclear magnetization.

Once the RF pulse is turned off, the net magnetization in an MR sequence will begin to recover toward equilibrium. The T1 relaxation time describes how quickly Mz increases back towards its equilibrium value, the longitudinal magnetization component. Mz will recover according to the exponential time course described by the differential equation

$$
\frac{dM_{z}}{dt}
=
\frac{M_0 - M_z}{T_1}.
$$

In a similar manner, the decay or loss of Mxy is described by the T2 relaxation time. After the RF pulse, Mxy, which serves as the source of the MR signal, will grow before decaying, following the equation

$$
M_{xy}(t) = M_{xy}(0)e^{-t/T_2}.
$$

The complete loss of transverse magnetization after a 90° excitation pulse is characterized by the T2* relaxation time, and reflects the loss of coherent precession. If there is any magnetic field inhomogeneity present during the spin echo time, the effective transverse relaxation time in the imaging slice becomes T2*. T2* is the quantity that is measured in the gradient echo sequence where no rephrasing of signals occurs [125, 126, 127].

Pulse sequences

Magnetic resonance (MR) imaging sequences manipulate the time-dependent nuclear magnetization of tissues within the radio frequency (RF) excitation and readout intervals. Two kinds of signal decay occur during MR imaging: transverse decay (T2*) and longitudinal recovery (T1). T2* decay is caused by interactions between spins in the same voxel and homogeneous B0 field inhomogeneities, while T1 recovery is determined by the environment (local lattice interactions) and the net magnetization at the point of excitation. The variations in T1 and T2 times in tissues create the contrast that makes MR images diagnostic. Non-phase-encoded, free-precession sequences are termed gradient echo sequences and are useful in many applications, including fast imaging.

The repetition time, TR, influences the contribution of T1 recovery to the signal. When TR is very short, T1 elements are large, but low signal-to-noise ratio (SNR) limits visualization. Conversely, with a TR long enough to attain full T1 recovery, the T1 element drops to zero and the signal falls to zero. Short TR (usually ≤ 2T1) increases SNR, but some signal drop occurs in fat. So much fat signal loss may be undesirable in abdominal imaging but can provide a unique T2-weighted imaging opportunity in musculoskeletal studies. Dark blood imaging, in which blood becomes hypointense, is another useful application. In fact, very short TR times (often as low as 1.5 TEs) improve imaging speed by decreasing the number of RF excitations before sampling (k-space temporal undersampling) [128, 129, 130].

Safety considerations (SAR, magnet hazards)

Electromagnetic radiation is produced by charged particles accelerated in a magnetic or electric field. Nuclear magnetic resonance techniques utilize a static magnetic field to generate nuclear magnetic resonance (NMR) signals based on electron trajectories.

Magnetic resonance imaging (MRI) exploits these NMR signals, harnessing the relaxation mechanisms of excited nuclei to create detailed tomographic images of biological tissues. Each nucleus equilibrates back to its equilibrium distribution and state according to its relaxation time constants: T1 for longitudinal relaxation, T2 for transverse relaxation, and T2* for dephasing induced by local field variations. A variety of pulse sequences can be applied to adapt contrast mechanisms; however, they all involve a trade-off between scan time, spatial resolution, and image signal-to-noise ratio.

Clinical MRI implementations can be hazardous due to the strong static magnetic field produced by the magnets. Ferromagnetic or susceptible materials close to the magnet can be projectile objects causing injury to patients or workers, while implants may be damaged or thermally affected by the field. Attention must also be given to high specific absorption rate levels, as deposited energy may produce tissue heating in susceptible regions or even provoke heating and burns in poorly perfused areas [131, 132, 133].

Chapter - 11

Nuclear Medicine Physics

Medical physics is a fascinating and multidisciplinary field where fundamental principles of physics are applied to the development and use of medical imaging, therapeutic devices, and clinical treatment of patients. In this area, physicists participate primarily as researchers in the advancement of new imaging and treatment methodologies and devices. Medical physicists also play a clinical role, ensuring equipment is properly calibrated, optimized, and functioning safely and effectively.

Clinical medical physics is primarily practiced within the areas of diagnostic imaging, radiation therapy planning and dosimetry, and radiation safety. In diagnostic imaging, the goals are to ensure that patient and occupational radiation doses are minimized—consistent with high image quality that meets the clinical requirements—through the design, optimization, and quality control of imaging equipment and systems. In radiation therapy, the primary responsibility is to establish the required dose distribution and determine how that distribution is delivered accurately and safely to the patient. For both areas, expertise is also applied in a consultative manner for emerging technologies and techniques that are only gradually becoming established procedures. Finally, through the application of radiation safety concepts and methodologies, the clinical medical physicist ensures that the risks to patients and staff from radiation exposures are ALARA (as low as reasonably achievable) [134, 135, 136].

Radionuclide production

Diagnostic and therapeutic radiopharmaceuticals normally contain short-lived radionuclides that are produced in small quantities and applied mainly in PET scans and SPECT images. In SPECT, radiopharmaceuticals are routinely administered to patients up to several hours before imaging to allow sufficient time for pharmacokinetics and are subsequently imaged using gamma cameras with SPECT-collimated detector systems. The most common radio-nuclide in clinical usage is Technetium-99m (Tc-99m), but often short-lived isotopes are produced and/or utilized in situ during an examination to minimize radiation dose to the patient. Typical examples include Gallium complexes and those involving Indium. In PET, positron-emitting radio-nuclides such as Carbon-11, Oxygen-15, Nitrogen-13 and Fluorine-18 are produced by cyclotrons. Generally, these isotopes have half-lives of less than 2 hours and so a mini-cyclotron is required on the premises of large hospitals or they are supplied by a local centre. The most widely used radiopharmaceutical is Fluorine-18 labelled Fluorodeoxyglucose (FDG) which is optically inactive and accumulates in metabolically active tissues.

The increasing demand for diagnostic imaging using radiopharmaceuticals and probes has led to the development of novel synthesizing modalities and new probes that enable more complex labelling of biomolecules. Radio-pharmaceuticals are also being developed that can selectively label specific cellular receptors, enabling multiple SPECT studies to be performed on the same patient, for example using probes labelled with the same isotopes 123I, 99mTc, and 111In. Radio-pharmaceuticals are also produced for organ-specific imaging or therapy in Gastro-enterology, Oncology and Rheumatology. Hospitals engage in research and development in collaboration with local universities and testing and production may even be sponsored by suppliers [137, 138, 139].

Gamma cameras and collimators

Gamma cameras are devices for imaging gamma radiation emitted from a radiotracer introduced into the patient in nuclear medicine studies. Although also known as scintillation cameras and gamma scintillation cameras, their conventional designation as gamma cameras now encompasses both older designs with scintillation crystals and more modern devices that employ semiconductor detectors.

Without collimators, the intrinsic resolution of a gamma camera is determined by the thickness of the scintillation layer and the quantity and arrangement of photoelectric cells coupled to the scintillator. In practice, however, thin scintillation crystals yield poor-quality images with inadequate contrast because of the overwhelming contribution of Compton-scattered photons. Collimators improve spatial resolution by limiting the angle of photons incident on the detector but at the cost of reduced sensitivity. Clinical images usually have pixel sizes of up to 5 × 5 mm2, with a minimum collimator hole size of 1-2 mm and a typical spacing of 3.5-5 mm.

Single-photon emission computed tomography (SPECT) employs a rotation of the gamma camera around the patient. Photon emission from the radiotracer is acquired from multiple angles, and the resultant projection data are reconstructed to provide transverse, coronal, and sagittal slices of the radiopharmaceutical distribution. Positron emission tomography (PET) exploits the annihilation radiation produced when radiolabeled molecules emitting positrons undergo transport and decay *in vivo* [140, 141, 142].

SPECT and PET imaging

Clinical images are commonly produced with X-ray photons. Nuclear medicine provides complementary imaging modalities: single-photon emission computed tomography (SPECT) and positron emission tomography (PET). In SPECT, γ rays are emitted from γ-emitting radionuclides that localize within the patient following intravenous administration of a suitable radiopharmaceutical distribution. The spatial distribution of γ-rays within the patient is recorded by a dedicated gamma camera and can subsequently be reconstructed to produce a 3D volume of radiopharmaceutical distribution. In PET, positron-emitting radionuclides are administered and the coincident detection of the resulting two 511-keV γ-rays allows for highly sensitive imaging of the spatial distribution of the radiopharmaceutical.

Radionuclide production differs between SPECT and PET. Many Technetium-99m (99mTc) radiopharmaceuticals are produced using a technetium generator system—commonly comprising ^99Mo (half-life 66 h)—which is positioned remote to the radiopharmacy. Other SPECT and PET radioisotopes are procured by irradiating a target material using a cyclotron, with the parent radionuclide often having a short half-life (e.g. I-124 half-life 4.18 d, Ga-68 half-life 68 m, Cu-64 half-life 12.7 h).

The use of radiopharmaceuticals that incorporate a γ-emitting isotope enables SPECT imaging. The radioisotope is produced in appropriate chemical form and the chemical structure of the radiopharmaceutical is designed to allow distribution characteristics that are suitable for clinical examination. The spatial distribution of the radiolabelled compound is determined primarily by the physiology of the target tissue. Despite the associated radiation dose, the total-body distribution of the compound provides information concerning functional processes in the body and may help in the differentiation between benign and malignant lesions [143, 144, 145, 146].

Image corrections (attenuation, scatter)

Corrections for attenuation and scatter are critical for quantitative accuracy in SPECT and PET. Attenuation corrections compensate for the reduction in signal intensity due to absorption and scattering as the emitted photons traverse the patient's body. In SPECT, the emission of lower-energy γ-rays increases the impact of attenuation, especially for isotopes such as 99mTc. While an analytical attenuation correction can be applied using a 2D map of the integrated attenuation coefficients along the projection rays, scattering and depth-of-interaction effects are more challenging. Although scatter can be corrected to a certain extent using empirical models, it persists in the final image and reduces the contrast-to-noise ratio. The impact of scatter on quantification depends on the processing methodology, the size of the organ of interest, and the isotope used.

In PET, the emitted 511 keV γ-photons have a shorter interaction length than those used in SPECT, permitting a more accurate analytical correction. However, the number of detected coincidences is not constant along the line of response, and extra events occur for two photons that interact in different tissues. Along with attenuation and scatter, partial-volume effects need to be considered for accurate quantification; the detected activity concentration is sensitive to the size of the region of interest and could differ significantly from the true concentration determined in a phantom. The combination of attenuation and scatter corrections enhances the quantitative accuracy of SPECT and PET, enabling measurements of tracer biodistribution with known doses [147, 148, 149].

Chapter - 12

Radiation Biology

Medical physics is usually considered as divided into two major branches: diagnostics and therapy. The distinction can be broad, with each branch further subdivided into a variety of specialty areas. The diagnostic branch is usually considered to encompass the production and processing of images within whose content information is available for a diagnostic purpose. The essential requirement of all diagnostic techniques is, of course, that they should yield images indicative of the functional or anatomical state of the patient, without causing any undue risk to the patient's health or safety. As a consequence, all radiological imaging systems require additional components, apart from the obvious image "sensor," to reduce the risk of radiation dose to the patient to a minimum generally considered suitable for the procedure being undertaken; this risk usually takes the form of extra radiation dose to sensitive tissues resulting from the scattering of radiation by the patient, producing secondary radiation that contributes to the image but without useful diagnostic information.

By contrast, the branch of medical physics dealing with therapy is concerned with the application of high-energy radiation for the purpose of destroying malignant or abnormal tissue. There is, however, an element of diagnosis associated with therapy; therapy guidance involves the positioning of the patient in the irradiated area and the recovery of images that assist in the definition of the shape and position of the beam. Despite such differences in these two broad areas, there exists a close relationship between the two branches. Diagnostic imaging was among the earliest applications of radiological science, and many of the essential concepts required for diagnostic imaging, including image quality measures and the validation of dose to the patient, were wholly or in part created within this specialty. It is also noteworthy that it was the early developments of diagnostic imaging that prompted the first regulations in the field, with special emphasis on patient safety [150, 151, 152].

Cellular response to radiation

Cellular responses to radiation are essentially those responses, at the cellular level, after dose delivery. Response may be classified as deterministic or stochastic, with the probable rates of occurrence of either effect varying with dose in an exposure situation. Deterministic effects are severity-related biological damage responses from macromolecular and cellular systems in excess of a known threshold dose, dependent upon the nature, dose, dose rate, and volume of tissue irradiated, while stochastic effects result from a complex interplay of factors and are generally considered to be non-threshold effects, the probability of occurrence being considered a function of dose rather than the severity. Primary responses in humans and other mammals are generally confined to the development of cancer and heritable effects, even if, at lower doses, deterministic tissue damage and even death following radiation exposures are possible.

The dose-effect relationships, sex differences, type of biological effect, and the type of radiation are needed to derive risk coefficients. Information about radiation risk has been compiled from a variety of patient data, research studies, accident data and, in some cases, from experimental radiobiological studies. Primary data sets on which risk coefficient estimates are based are the Life Span Study study of atomic bomb survivors and the medical studies of patients exposed to therapeutic radiation. Life Span Study risk estimates, which represent the first epidemiological study in irradiated people, have contributed most to the entire picture of radiation-induced cancer risk and have been extended, among other aspects, to consider the effect of small and protracted doses in the low-exposure range.

Risk estimates for solid cancer and leukaemia can be applied to other exposed populations, such as those who participated in the early nuclear weapons test programs in the Marshall Islands or elsewhere. Information on radiation heritable effects in humans, derived principally from off springs of irradiated mice and dwarfs, is summed and the effects have been included in human risk assessments with downward adjustments based on cautions arising from the imperfect knowledge of the underlying mechanisms involved. Nevertheless, the assessment of risk and possible radiation damage to the exposed population remains a high priority in the collaborative work of ICRP, UNSCEAR, and other organizations involved in documenting biological and health effects of radiation exposure [153, 154, 155].

Deterministic vs. stochastic effects

Advances in cellular and molecular biology have revealed that in a simplistic way, cells respond to radiation in 2 different ways: following large doses (above few 10's of centigrays) cell membranes are severely damaged and it is apparent that most of the cells die before dividing. This is called a deterministic effect and in clinical applications these effects are known as tissue reactions. However, below these threshold doses, the concept of radiation protecting a human as a realistic population is not valid. The prevailing theory is that radiation hazard depends on genetic material and is the result of very rare and accidental events. Measurements of probability of these events are estimated in human studies based on the exposure and follow-up of such population and the prediction of the expected incidence of these diseases based upon the natural data (dose-response relationship). Therefore the predicted excess incidence rate is defined as stochastic effect at the level of an individual. What differs the 2 effects is the dosage range described: safety limits are based upon the stochastic effects.

Risk assessment of health detriment as a result of exposure to radiation doses is one of the most challenging areas of radiation protection and risk assessment involves establishing a dose-response relationship, that is, a mathematical description linking radiation effects and exposure dose. Laboratory workers subsequently discussed estimating dose-response relationships for a wide variety of human diseases, using various populations under different conditions. Over the past 50 years, risk assessments have been performed for both radiation-exposed workers and members of the general public. The creation of a cancer in an exposed person is regarded as an incident based on a genetic accident and the statistical probability of producing such an accident is related to the external dose, expressed as a risk. Reports on Health Effects of Low-level Radiation expose the concept of risk and clearly state in terms of population groups [156, 157, 158].

Dose-response relationships

Dose-response relationships describe how specific biological responses vary with radiation dose and are critical for understanding deterministic and stochastic effects. Deterministic effects (e.g., tissue injury, organ dysfunction) occur when the absorbed dose exceeds a threshold, typically determining local treatment tolerance. Stochastic effects (e.g., cancer, genetic anomalies) are random in nature but become increasingly probable with rising dose. The relationship between dose and response typically follows the form of a first-order exponential: the excess probability is represented by a dose-proportional term, whereas all residual risks are captured by the intercept.

Dose-response relationships inform the clinical risk assessment that underpins radiation protection. Fatal cancer estimates are derived from survivor studies and serve as a basis for calculating cancer risks in patients undergoing diagnostic imaging. Epidemiological investigation of abnormal births has provided a foundation for estimating the hereditary risk of irradiation. Consideration of the shape of the dose-response data has brought into question the use of probability factors that remain valid over a wide range of doses, and it has stimulated discussions about exposure reduction in screening programmes.

Radiation risk assessment

Translates knowledge of deterministic and stochastic effects into clinical practice. Clinical exposures involve low doses, primarily of X- or γ-rays, to highly radiosensitive tissues. Risk models derive from cell populations in animal models, augmented by epidemiological studies of Japanese atomic bomb survivors, radium watch-dial workers, and several medically irradiated cohorts. Current analyses generally support the linear-no-threshold model. The approach is conservative, reflecting an ethical duty of care to patients; cancer induction is considered possible but not quantifiable.

Probabilistic models for radiation-induced cancer and heritable disease have led to risk estimates that influence medical radiation safety. Risk estimates of excess cancer deaths are based on A-bomb survivor data. Quantitative cancer risk models typically combine a radiation dose-response and a background rate function. Sources inferred from *in vivo* experiments require consideration of organ differences in radiation response and energy absorption. Organ doses, and, hence, risks, are determined from radiopharmaceutical biokinetics using biokinetic models validated with human data. Risk estimates of cancer induction thus incorporate cancer-prone sites, informed by radiation-induced cancer studies, and internal doses based on biokinetic models validated against human data [158, 159, 160, 158, 159, 160].

Chapter - 13

Principles of Radiation Protection

In Diagnostic Physics and Therapeutic Physics, the medical physicist strives to achieve three goals: first, to conduct examinations and any type of procedures that involve radiation in a way that yields the required information at the lowest possible risk and injury to the patient and healthcare staff; second, to make possible the treatment of cancerous and other pathological tissues in a way that destroys them or hinders their development without excessive injury to surrounding healthy tissue; and third, to ensure that all equipment used in radiation-based examinations and treatments operates safely and optimally in accordance with predefined specifications.

In most countries, the medical physics practice is regulated and accreditation bodies require a medical physicist qualified in the specific field of practice for the equipment involved. For diagnostic applications, this is in accordance with separate documents and recommendations related to ionizing radiation exposure in diagnostic radiology (X-rays, radiological and nuclear medicine), fluoroscopy, interventional radiology, mammography control, and dental (as appropriate) applications; for therapeutic applications, it involves assuring the safety and proper configuration of equipment for external-beam therapy (linear accelerators, cobalt treatments, brachytherapy) and brachytherapy delivery systems (afterloaders, applicators) [161, 162, 163].

ALARA concept

The ALARA (As Low as Reasonably Achievable) principle denotes a general approach to radiation protection and is upheld universally. Its implementation requires the adoption of appropriate practices, including radiation shielding design. Practice-specific recommendations concerning ALARA and radiation protection of workers are based on recommendations from the ICRP and IAEA. Medical physicists bear a key responsibility for implementing the ALARA principle in radiology and radiation therapy. Personal dosimetry assesses individual exposure and is required for selected workers likely to exceed set exposure levels.

Radiological protection is an important aspect of many imaging and therapy modalities. Radiation risks are not negligible, and patient and public safety remains a high priority in medicine. Consequently, risk mitigation is a priority in all imaging and therapy applications, and specific criteria have been established for each modality. Apart from potential exposure arising from specific procedures or treatments, such as CT scans, all controlling authorities have implemented a dose-monitoring framework. In addition, all monitoring procedures are regularly evaluated to ensure continuous decreases in patient doses. Special factors to be considered include justification and optimization. Justification aims to ensure that a procedure's benefit far outweighs the risk, while optimization aims to keep the doses as low as reasonably achievable, taking into account social and economic factors. While justification lies primarily with the referring physician, the responsibility for optimization typically falls on the medical physicist, who must maintain and operate the selected equipment efficiently, provide appropriate and correct protocols, and use good practice in all procedures [164, 165, 166].

Shielding design

In radiation protection, shielding is one of the main methods to reduce exposure of workers and the general population to ionizing radiation. The functioning of the shielding is a result of the interaction of the radiation with matter, as described in Section 3. Therefore, before showing the approach used for shielding calculations in medical physics, it is useful to cite the attenuation law for monoenergetic radiation:

$$
I = I_0 e^{-\mu x}
$$

\]

where

- I is the intensity transmitted through a thickness x of material,

- I_0 is the intensity without material,

- μ is the linear attenuation coefficient, characteristic of the material and the radiation.

As shown in Section 3.4, μ varies with radiation type and energy. The shielding design must follow the ALARA principle while considering the specific characteristics of the institution, equipment, and procedures. Special attention is needed for therapists and first responders in locations of high dose. The required thickness of the shielding is proportional to the dose rate outside the shield and inversely proportional to the attenuation properties of the absorbing material. Moreover, the shielding design must follow the prerequisites of ICRP Reports, the BSS (Basic Safety Standards), and local regulations.

The selection of the proper material and its thickness is made taking into account the anticipated dose rates from radiological equipment or the radioactive sources to be managed. Informative data may be obtained from shielding tables that classify adequate materials for the different types of radiation. Thus, for nearlly all sources, some form of hydrogenous concrete will provide both fast neutron attenuation and gamma protection. For high-energy gamma sources, dense materials such as lead or depleted uranium [167, 168, 169].

Personal dosimetry

Encompasses the monitoring of occupational radiation exposure for personnel engaged in diagnostics and therapy involving ionizing radiation. Research indicates that the majority of radiation exposure to medical staff originates from scattered radiation during procedures like interventional radiology, particularly in fluoroscopy and cardiology. Given the relative magnitudes of effective dose from imaging and therapeutic applications, conventional exposure limits for both medical staff and patients necessitate continuous assessment.

Monitoring systems typically consist of a detector and electronics coupled with display and processing software. The most commonly used devices for personal detection include film dosimeters, thermoluminescent dosimeters (TLDs), optically stimulated luminescence detectors (OSLDs), personal electronic dosimeters, and radiophotoluminescent dosimeters (RPLDs). For high-energy photon and beta radiation, ionization detectors serve as reference instruments in clinical dosimetry, air kerma calibration, and the establishment of clinical protocols. In the low-energy range of radiation, TLDs, OSLDs, and RPLDs have gained preference and are extensively applied in neutron dosimetry due to their compact dimensions. The ICRP recommends reporting for personnel dosimetry measurements in two ways: against dose limits and against the dose distribution from TL and OSLD detectors [170, 171, 172].

Occupational exposure limits

All countries impose limits on occupational exposure to ionizing radiation for personnel involved in the use of radiation in medicine. While these limits are typically derived from international recommendations, they are enshrined in national legislation. Radiation safety audits usually include assessment of compliance with applicable laws and regulations.

Are expressed as annual doses, usually within an appropriate nested set of equations that serve military, research, aircraft and frequent flyer, and medical groups. For the comprehensive group working in radiation areas, exposure to the whole body must not exceed 20 mSv/y, averaged over five years with a maximum of 50 mSv in any single year, 500 mSv in the lens, 150 mSv in the skin and extremities, and 1,000 mSv for tissues and organs other than the lens.

Although most exposure limits are unlikely to result in acute health effects from deterministic responses, the warning to limit high doses to the lens is an important element of the prevention of radiation cataracts. Guidance levels for nuclear medicine workers have been developed to ensure that doses remain below levels associated with an appreciable cancer risk. Dose monitoring programs are implemented in hospitals and clinics performing these procedures [173, 174, 175].

Chapter - 14

External Beam Radiation Therapy Fundamentals

Medical physics has multiple branches:— In diagnostic medical physics, application of physics concepts, principles and methods to ascertain the suitability of a radiation source for a diagnosis or to obtain a diagnostic image of sufficient quality for interpretation with regard to the anatomy, physiology or pathology of the imaged region. Diagnostic medical physics includes the use of ionizing and non-ionizing radiation, including X rays, gamma rays, radioactivity, magnetic fields, light and ultrasound.

— In therapeutic medical physics, application of physics concepts, principles and methods to plan and implement a radiation treatment for neoplasia or other diseases that have a specific response to radiation exposure. Therapeutic medical physics encompasses distinct modalities of radiation treatment, namely brachytherapy and external beam therapy, and their combinations. External beam therapy employs X rays, photon beams from medical accelerators, and mainly high-energy neutron, proton and heavy-ion beams.

— The work of the medical physicist includes ensuring that the optimum quality of service is provided by managing and overseeing the physics aspects of the installation, commissioning and clinical use of equipment, and the physical aspects of radioprotection. The latter incorporates determining the diagnostic reference levels and the respective control, and managing the personal dosimetry of employees requiring the wearing of, and hence the possibly excessive radiation exposure shown by, dosimeters [176, 177, 178, 179].

Linear accelerators

Modern external beam radiation therapy makes extensive use of linear accelerators (linacs), with cocentral photon and electron beams bein-g employed for a majority of treatment procedures. The X-ray beams generated in linacs have energy spectra Pennitting the simultaneous treatment of targets of different depths, while electron beams can be used for superficial tumours, with a rapid drop of dose beyond the treatment volume. In addition to the primary beam, which is used to irradiate the patient, linacs are provided with various beam modifiers, including collimators and bolus material. The treatment head contains the components that directly or indirectly shape the clinical beam and influence its dosimetric characteristics.

A linear accelerator is an electromechanical device, designed to generate and direct high-energy particle beams for a variety of applications in basic and applied science, industry or medicine. In radiation therapy, these devices are used to produce and deliver highenergy X-rays or electrons for the treatment of cancerous tumours. A basic linear accelerator designed for medical applications is shown in Fig. 14.1. In the conventional mode of operation, the accelerated electrons are directed onto a target to produce a therapeutic X-ray beam. Special beam-shaping and filtering devices are incorporated in the treatment head to optimize the characteristics of the clinical beam for patient treatments. Any such device serving a therapeutic function can be regarded as a treatment head. Specialized systems incorporate additional radiation sources, either for enhancing the properties of the primary beam or for providing a second, auxiliary beam of different characteristics. These systems commonly employ high-energy photons or electrons [180, 181, 182, 183].

Photon and electron beams

Antimatter particles are seldom present in nature, yet positrons and electrons consistently condense under radiation fields. The intense acceleration fields formed in high-energy radiotherapy systems permit particle annihilation by pair production, resulting in the emission of two neutrinos in opposite directions. Other pair-production interactions yield positron-electron couples propagating through matter until they recombine into a photon of 0.511 MeV. The resultant photons subsequently transfer energy to matter in a variety of ways, with the particle ceasing to be significant in developing areas such as boron-neutron capture therapy. Neutron therapy systems exploit significant nuclear cross-sections to localise energy deposition.

Are fundamental radiation sources utilized in medical diagnostics, surgery, and therapy. The continuous emission of bremsstrahlung radiation originates from closely bound electrons, and characteristic radiation arises through ionisation-produced disturbances in excitation equilibrium. Photon beams emitted from a point source produce images in projection radiography at the end of a long path. The tube target and anode of X-ray systems scan radiographs and fluoroscopic images. For radiotherapy treatment planning, the quality of the deposited dose is referenced to depths in the Bragg peak, beyond which fast electrons cease to contribute. Hence, both beam quality and beam data characterising dropout are necessary.

Large-scale conical and cylindrical neutron-producing targets generate particles primarily for irradiation experiments in biology and radiation damage investigations and for potential clinical applications in boron-neutron capture and fast-neutron therapy. Particle annihilation also creates soft X-rays suitable for therapy. Cyclotrons produce particles of selected energies for transmission and for the manufacture of radionuclides and radiopharmaceuticals. Nuclear radiation applied as a vector may also be shaped into radiological, nuclear, or sounding weapons. The clinical or radiological characteristics and the history of use of medical sources, together with the relevant protection and radiation hygiene aspects considered in the respective sections for patients and personnel, determine their roles and suitability in all applications [184, 185, 186, 187].

Beam modifiers

Comprise a diverse class of devices incorporated into or mounted on a linear accelerator treatment head to reshape the radiation distribution of a therapeutic treatment beam in five principal ways: flattening, boosting dose in specific regions, modifying relative biological effect, modifying energy or spectrum, and shielding. In addition to bolus material, collimators that produce irregular treatment fields are also beam modifiers, as are asymmetric beam collimation systems, beam spoiler systems, and devices that produce compensating beams.

The most widely employed, and technically the simplest, beam modifier is the flattening filter (FF). The primary function of an FF is to produce a radiation beam exhibiting a relatively uniform dose distribution over a rectangular field. This is achieved by placing a filter material with a gradient of high-to-low attenuation in the beam axis. The theory of FF design has been reviewed recently. The basic design criteria are confined principally to beam energy and the desired degree of flatness. The selection of filter material depends on the dose rate of the clinical accelerator: a high-Z material is usually used for dose rates greater than 1000 cGy/min and a medium-Z material for lower dose rates.

The relative effectiveness (RBE) of neutron radiation is significantly higher than that of γ-rays and X-rays for the same absorbed dose. Radiotherapy using neutrons or any mixed beam that contains neutrons requires a method of measurement and reporting of the absorbed dose that accounts for this effect. A quality factor (Q) has been introduced for this purpose, which provides the ratio of the absorbed dose from neutrons to the absorbed dose from conventional γ-ray or X-ray radiotherapy that would produce approximately the same deterministic effect, as defined in Section 12.1. The neutron QF is unity ($Q = 1$) for $\rho > 10$ MeV, and increases for lower energies, tending towards a value of 30 for thermal neutrons.

In addition to the principle of relative biological effectiveness, a treatment plan incorporating neutron-beam therapy should include consideration of dose-differential response factor (QF). The QF serves to estimate the absorbed dose in the sensitive organs or tissues along the neutron-path when irradiating with a mixed beam containing both neutrons and the high-energy photons of cobalt, cesium, or any clinically used weight gain source, but is valid only when the absorbed dose of the neutrons in the same sensitive part is measurable [188, 189, 190].

Treatment head components

The treatment head of a linear accelerator supports source shielding, distributes the radiation beam, and incorporates mechanisms for enhancing the delivery of prescribed dose distributions, usually involving the deployment of collimators, beamsplitters, filters, and bolus.

The most abundant beam modality is a high-energy photon beam produced by a linear accelerator. Its treatment head performs the following functions:

1. Shielding of the radiation source.
2. Shaping of the beam distribution with the aim of matching the clinical need.
3. Controlling the clinical dose distribution through the use of modulators.
4. Reducing the radiation dose to personnel in the vicinity of the patient.
5. Generating KV beams for imaging purpose.
6. Defining the isocentre of the machine by the use of beamsplitters.

The main components of the treatment head are:

1. The micro-multileaf collimator (MLC), which defines the sharp lateral edges of the treatment fields and has the capability of making “non-convex” fields.
2. The primary collimator, which shapes the clinically needed beam, regulates the beam divergence, and controls the radiation dose outside the treatment fields.
3. The secondary collimators, which are used to select the radiation size at source distance and can provide additional attenuation in the beam.
4. Total filter systems, which remove unwanted low-energy radiation that contributes to patient dose without a therapeutic benefit.

Chapter - 15

Radiation Therapy Dosimetry

Medical physics is defined universally as the application of physics to medicine. However, within the field of medical physics there are differences in applied physics specialities, broadly concerning either radiation diagnosis or radiation therapy of patients. These specialities have historically evolved separately, with different legislative regulations and safety standards; different accreditation and Department of Health Inspectorate requirements; and, to some extent, different principles of practice. In radiation diagnosis these specialities have also been labelled `the physics of radiation production, the physics of image formation and display and the physics of radiation detection'.

These specialities may be more clearly defined by reference to areas of clinical application in radiation diagnosis: the production and detection of either diagnostic or therapeutic radiation; the processes by which images are produced from radiation intensities recorded by detection systems; and the properties of systems that detect, but generally do not record, radiation directly, such as image-intensifier systems [191, 90, 192].

Reference dosimetry protocols (AAPM, IAEA)

Reference dosimetry for photon and electron beams in radiotherapy is performed according to protocols established by AAPM Task Group 51 and IAEA TECDOC-1583. IAEA protocols for clinical use of high-energy photons in brachytherapy, clinical dosimetry of high-energy photon and electron beams, and reference dosimetry of high-energy photon beams are also available. AAPM and IAEA protocols cover site-specific cavity theory, the requirement that reference conditions are satisfied for a water phantom, and that conditions for the determination of output factor data are valid.

Reference conditions involve the use of an ionization chamber calibrated in cobalt-60 radiation with the following characteristics: (i) Pion, (ii) barrier-free design, (iii) radial placement for electron irradiation (IEC 60731 designation A type), (iv) vented, and (v) usable in beams with |S| < 0.1 for cobalt-60 to cesium-137. Measurements are performed in a well-defined water phantom by a qualified medical physicist and are closely supervised by the radiation oncologist. The values are used for patient-related dosimetry, dose calibration, and for monitoring the radiation treatment delivery quality [193, 194, 195].

Output factor determination

Output factors (OFs) quantify treatment head output as a function of field size, beam angle, geometry, and gap between the treatment head and the phantom surface. Measured reference output factors are defined for a specified combination of monitor units (MU), linac type, treatment head, water-equivalent phantom, and reference arrangement. Output factors can be expressed as ratios of output for arbitrary conditions to the reference value, reducing sensitivity to systematic and random variation.

An output factor measurement is performed when all uncertainties are at acceptable levels, preferably in conjunction with a commissioning process consistent with the recommendations of the AAPM and IAEA. The detectors best suited for output factor determination vary based on the specific output factor being measured. For small field sizes, although solid-state detectors are preferred, it is common practice to use ionization chambers. Large fields typically require either moving aquapans or dose-rate compensating detectors. Photomultiplier-based detectors with associated water phantoms are useful in measuring output factors at other energies [196, 197, 198].

Beam data commissioning

Represents a vital step in treatment planning and dose delivery for external beam radiation therapy. Treatment beams must be carefully characterised: their absolute output and the variation in beam profiles and depth dose distributions with field size are measured, as are the factors necessary to account for the impact of beam modifiers, including physical wedges. These data are used by treatment planning systems—software solutions equipped with dose calculation algorithms—to generate and optimise patient-specific treatment plans. Accurate commissioning is the prerequisite for reliable treatment delivery, periodic checks are performed to confirm the continuing integrity of the beam data.

In addition to patient plan verification, in-vivo dosimetry provides an independent means of checking the delivered dose. Specially designed dosimetry devices are used to measure the dose delivered to a patient, either during the treatment or in a subsequent CT scan. The results are compared with the predicted dose based on the treatment plan to check for systematic errors, allowing for rapid adoption of corrective measures when required [199, 200, 201].

***In vivo* dosimetry**

In external beam radiotherapy consists of dosimetric measurements in patients for one or more beam energies, using a beam quality for which a reference dosimetry protocol has been established. The purpose of in-vivo dosimetry is to detect major treatment delivery errors and to quantify delivered doses. As a quality-assurance tool, its importance lies not only in the evaluation of the delivered dose to the patient, but also in identifying the probability of major mistakes in dose delivery. These mistakes can be caused by incorrect electronic data input, wrong patient position, machine failures not detected by routine checks, and incorrect dosimetry calculations.

The main assumption is that both the in-vivo detector and the simulation of its response can be considered as ratiometric measurements and that the ratio is essentially independent of the patient-specific conditions. This is achieved by measuring not only for a normal clinical setup but also for a number of various special conditions, such as field misalignment, patient weight changes, lung compensation, and in-vivo measurements with the beam collimated to 1 × 1 cm2 [202, 203, 204].

Chapter - 16

Treatment Planning in Radiotherapy

Medical physicists are specialized scientists trained to apply the principles and methodologies of physics to medicine. Their main contributions are in the fields of diagnostic and therapeutic medical physics and radiation safety. In medical diagnostics, medical physicists are responsible for the acquisition of quality medical images using the minimum possible radiation dose. In medical therapy, they plan and monitor patient treatment with radiation. Medical physicists pay particular attention to the assessment and minimization of radiation risks for patients and medical staff. In addition, they offer scientific supervision and expertise in the production of radiopharmaceuticals used in diagnostic and therapeutic nuclear medicine.

A coherent group of professional and regulatory bodies such as the International Atomic Energy Agency, the American Association of Physicists in Medicine, and the European Federation of Organizations for Medical Physics integrates recommendations from scientific research into regulatory frameworks that cover all aspects of medical physics. These also include ethical directives pertaining to the profession. Thus, accredited clinical and academic medical physicists can provide reliable clinical service to patients and at the same time make a significant contribution to the further development of their field [205, 1, 206].

3D-CRT, IMRT, and VMAT principles

To combine the advantages of 3D conformal radiotherapy (3D-CRT) and intensity-modulated radiotherapy (IMRT), volumetric modulated arc therapy (VMAT) applies the principles of intensity modulation and transverse image-plane photometric beam-shaping to treatment delivery, with the beam continuously rotating around the patient. Overlapping sections of the arc can be allowed for more complete modulation. The mechanical requirements and constraints, and the considerations for imaging during treatment, can differ from those in conventional kV or MV imaging.

The delivery process generally begins with planning the treatment trajectory, either on a dedicated treatment-planning system (TPS) or directly on the treatment-planning system of the Linac using a geometrical approach. The former aims to split the arc into sections of low modulation, which can be treated by a regular IMRT algorithm. Direct planning in the treatment-planning system distributes the dose across the arc, but requires additional testing of the planned trajectory for excessive lead-angle modulation.

The angle for each section of the arc is determined by the photometric state of the beam and the absolute request. The photometric state is expressed by the dose-Distance-to-Source relationship of the beam. The photometric requirements are always fulfilled, but its dependence on the isodose positioning determines the dose delivered to the patient.

Dose calculation algorithms

Modern treatment planning systems employ sophisticated and accurate dose calculation algorithms to determine the spatial distribution of radiation dose in the patient. The resulting dose distribution is then evaluated using an array of clinical criteria (see Section 15.3) and is used to make a decision regarding the treatment plan to be delivered. The success of the subsequent treatment depends on how the absorbed dose distribution of the actual irradiation compares to the predicted dose volume (or other relevant criteria).

The dose-calculation problem is a complex one, requiring consideration not just of the physical aspects of radiation dose deposition, biological factors such as dose-volume-histograms (DVH) and equivalent uniform dose (EUD) but also operational matters such as the availability and expected use of treatment modalities. Treatment planning systems approach the dose-calculation problem in different ways but can be classified as either convolution/superposition-based or Monte Carlo-based algorithms. The most commonly used methods are briefly introduced here [207, 208, 209].

Contouring and margins

Pre- and post-treatment contouring is a crucial aspect of radiotherapy treatment planning, affecting both the robust delivery of the prescribed radiation dose to the target and the minimization of dose to nearby healthy organs. From a clinical perspective, contours and the associated margin definitions ensure that all treatment scenarios—elaborate treatment plans, quality assurance verifications, etc.—accurately reflect the patient's anatomy and treatment approach. Contouring techniques involve a complex interplay of experience, clinical judgment, and advanced anatomical knowledge.

Inherent in the distribution of the prescribed radiation dose is uncertainty arising from patient and setup geometry or treatment delivery techniques; therefore, it is typically infeasible to directly irradiate the clinical target volume (CTV) to the prescribed dose. Instead, the dose is delivered to a larger planning target volume (PTV) that encompasses the CTV while accounting for these uncertainties [210, 211, 212].

Plan evaluation metrics (DVH, conformity index)

In radiation therapy treatment planning, dose-volume histogram (DVH) evaluation, whose importance has increased in parallel with the widespread implementation of treatment planning systems, facilitates analysis and comparison among anterior treatment approaches. Composed of dose-volume data on the target and critical structures involved in the plan, these graphical depictions convey the distribution of absorbed dose as a function of the volume of a specified structure. Among the treatment evaluation metrics derived from the DVH, the conformity index has gained increasing prominence in conjunction with the adoption of intensity-modulated radiation therapy (IMRT) and volumetric-modulated arc therapy (VMAT) techniques, in which the precision of dose delivery has vastly improved.

Developed to compare IMRT and VMAT against three-dimensional conformal radiation therapy (3D-CRT) and assess treatment plans with respect to target and critical structure doses, the conformity index quantitatively enumerates the degree to which the treatment is delivered to the intended target with minimal delivery to surrounding tissues. Ranging from 0 to 1, higher conformity index values are more clinically favorable. The inclusion of DVH-based metrics in the evolving planning and quality assurance toolbox expedites decision-making regarding treatment approach and practice, alongside the remaining well-established schematic and analytical methods for plan evaluation and comparison [213, 214, 215].

Chapter - 17

Brachytherapy Physics

The development of diagnostic and therapeutic physics has followed two independent paths throughout most of its history. Diagnostic physics began with the discovery of X-rays by Wilhelm Röntgen in 1895, while radiation therapy physics emerged shortly thereafter with the therapeutic applications of radium by Pierre and Marie Curie. Over the last few decades, advances in diagnostic imaging technology in particular have further brought these two fields together, resulting in a shift in emphasis towards medical imaging and patient safety. Despite the diversification of medical imaging modalities and the specialization required for their operation and interpretation, the underlying principles of medical imaging and exposure remain the same, and are therefore well understood by clinical medical physicists.

At the same time, the evolution of therapeutic radiology physics and radiobiology has established a framework that enables an assessment of the clinical effects of imaging techniques and their associated radiation dose. Furthermore, accreditation standards, regulatory requirements, and the principle of optimization all drive hospitals and medical clinics to keep the radiation exposure of patients as low as reasonably achievable (ALARA) without compromising diagnostic quality. The values of radiation exposure for diagnostic procedures are in general one or more orders of magnitude lower than those associated with therapeutic applications. Nevertheless, sufficient attention toward clinical practice should minimize the health risks related to exposure [90, 216, 217].

Radioactive sources and isotopes

Radioactive sources used in medical applications are classified according to the range of energy levels. Sources emitting radiation with a range of hundreds of keV or greater are usually produced in a nuclear reactor, while radionuclides with energies in the MeV range come from either high-energy accelerator facilities or laboratory synchrotrons. Radionuclides formed as a result of natural radioactivity or the burning of U, Th, and Pu fuel in reactors usually emit radiation in the MeV range.

The following types of radionuclides can be employed in medicine: brachytherapy sources, positron-emitting radiopharmaceuticals used in positron emission tomography, and radiopharmaceuticals for gamma cameras based on single-photon emission computed tomography. A special case of a radioactive source is a radioactive calibration source, which is used in hospitals and radiological laboratories for the calibration of radiation-dosage units [218, 219, 220].

Applicators and delivery systems

The most commonly employed isotopes for high-dose-rate (HDR) brachytherapy are 192Ir and 60Co, while low-dose-rate (LDR) brachytherapy employs isotopes with low neutron-gamma partition ratios such as 137Cs and 103Pd. Delivery systems comprise a high-activity radiation source housing which is withdrawn by remote control in a catheterized applicator, the position of the source being registered in real time by an indication display. The applicator is carefully designed to ensure an accurate reproducible position of the source with respect to the tumor. Care is taken to avoid radiation damage to cells during source placement and after treatment. The geometrical arrangement is to ensure dose equivalence at the surface of a specific applicator-tumor combination. In volumetric brachytherapy, the activity of the sources and their distribution are appropriately controlled for the region of interest, the specifics being determined according to the type of radiation source and the geometry of the target. The position of each source is recorded in such a manner that accurate reproduction is assured, and the arrangement is designed to enable rapid placement with minimal cell damage.

For all the afore-described treatment types, special applicators and insertion devices have been developed to ensure clinical practicality. The efficiency of dose distribution in brachytherapy is critically dependent on surface and excised dose rates. Such evaluation of neoplastic tissue is particularly important in HDR therapy because the large gradient in dose distribution can approximate surface dose-rate equivalence. Reasonable dose-rate equivalence also exists in post-surgical depth-dose distributions for the majority of clinical procedures. HDR techniques have recently been proposed in which a residual dose is depicted by an external beam source. Properly designed surface applicators afford a substantial reduction in corneal exposure ratios [221, 222, 223].

Dose distribution characteristics

The dose distribution characteristics and clinical consequences of high-dose-rate (HDR) and low-dose-rate (LDR) brachytherapy are profoundly dissimilar. Highly radioactive HDR sources are rapidly moved in and out of place to deliver high doses over a very short period, resulting in very high dose gradients around the source position, while LDR sources emit low activity, low-energy radiation, remain at the treatment site, and result in limited dose gradients. LDR sources are used in mucosal disease in the gynaecological and oral cavity regions and prophylactically in certain breast cancer treatment regimens. HDR brachytherapy is widely used in treatment of gynaecological cancer with other sites being considered. There is considerable interest in using HDR for interstitial skin cancers.

Treatment planning in HDR brachytherapy incorporates existing clinical data as opposed to personalised values. Poorly defined parameters such as a ratio of dose delivered at 2 cm from the surface with respect to the dose at the surface and treatment time with respect to volume of tissue treated have been used for LDR. These observations have fostered the use of LDR in superficial treatments although not based on direct evidence. The surface dose for LDR is much lower than that for LDR phosphorus implants resulting in less effect on the skin with LDR. The recommended criteria for LDR again have the limitations of being based on poor or even conflicting data. Doses for normal tissue contain a large uncertainity and normal tissue tolerances for all radiation methods [224, 225, 226].

HDR vs. LDR brachytherapy

Radioactive sources used in brachytherapy are classified as high-dose-rate (HDR) and low-dose-rate (LDR) sources. HDR sources are small, typically <1.8 mm in diameter, and deliver a high dose (>0.5 Gy/min) in a short time (<30 minutes). They are implanted temporarily via stepper or after loader devices. LDR sources are larger, can be adherent or non-adherent, and continuously impart a low dose (≤0.1 Gy/h) over a few hours, days, or weeks. Due to their small size and placement in multiple channels, HDR sources can provide patient-friendly dimensionality and precision.

Cueing a point-source approximation at a given distance "d" in a medium with linear attenuation coefficient (μ), the percentage depth dose (PDD) expresses variation in dose fallen onto the tissue. In contrast, the dose distribution of LDR sources has tabletop symmetry, resulting in negligible PDD variations. HDR brachytherapy has become viable for treatable sites using a single or multiple applicator catheter configuration, given inherent risks like seed migration, patient misplacement, and dose fall-off nature. The comparable dose distribution characteristics of HDR and LDR with appropriate normalization support clinical acceptance [227, 228, 229].

Chapter - 18

Image-Guided Radiotherapy (IGRT)

Besides basic principles and methods of medical physics, practical knowledge and skills are needed to ensure a safe clinical operation for X-ray diagnostic imaging and image-guided radiotherapy. Medical physicists contribute to the technical performance of devices and the quality of patient and operator safety by participating in tests, inspections, and calibrations undertaken during the clinical use of the devices. Such contributions are required by national regulations and international guidance documents promulgated by the International Atomic Energy Agency and the International Commission on Radiological Protection. They are also specified by national and international accreditation bodies, such as the US-based American College of Radiology. These radiological organizations expect clinical medical physicists, radiation protection officers, and radiological technologists to meet their respective obligations as established in the AAPM-ACR SF 1a publication, while the relevant clinical medical physics and radiation protection practices are further defined in the AAPM-ICRP joint publication, which describes the principles, requirements, and procedures that ensure compliance with local regulations and accredited practice.

Projection radiography, fluoroscopy, ultrasound, computed tomography, magnetic resonance imaging, nuclear medicine, and radionuclide production are routinely used diagnosis procedures. Safety and quality aspects in imaging associated with radiation dose reduction and image quality optimization are major areas of involvement for clinical medical physicists, who supervise these modalities also in many other specialty areas, such as pediatric, trauma, and interventional radiology, where different procedures, protocols, and techniques are implemented. Their clinical need and activities for using imaging in therapy guidance have increased with the development and overuse of computed tomography and computed tomography-fluoroscopy for image-guidance during treatment delivery. The magnetic resonance imaging-capability for on-line imaging incorporated in magnetic resonance imaging-linac systems represents another important complementary application [230, 134, 231].

kV and MV imaging

Radiation therapy imaging systems present specific technical and clinical requirements that differ from those of radiographic or fluoroscopic imaging. Kilovolt (kV) and megavolt (MV) imaging modalities—system types employed for treatment verification—are also characterized by lower image quality and higher radiation exposure than standard radiology or fluoroscopy systems. The need for accurate information pertaining to patient setup, bolus region definition, and treatment assistance has resulted in the development of surface imaging techniques and cone-beam computed tomography (CBCT). Although offering better image quality and lower dose in comparison with treatment imaging, these systems are not as commonly employed as kV or MV imaging systems.

Kilovolt (kV) and megavolt (MV) imaging are commonly used to verify patient setup, define bolus regions, and assist in treatment delivery. Treatment imaging, while not often included in the imaging protocol, is usually characterized by lower image quality than that of conventional radiology or fluoroscopy and higher exposure levels due to the anticipated short acquisition times. Radiation safety and patient dose concern factors such as efficient and proper selection of imaging modality, contrast media, and image enhancement procedures [232, 233, 234].

Cone-beam CT

(CBCT) employs volumetric data acquisition with a C-arm X-ray system, and image reconstruction algorithms produce 3D datasets. CBCT is used mainly in interventional radiology as a therapy guidance tool; however, its use can be extended to non-fluoroscopic examination methods. The primary imaging interest is bone and dense structures, resulting in lower dose per radiographic view when compared to routine fluoroscopy and CT.

Unlike conventional projection radiography, tissue volume can be imaged in multiple orientations with a C-arm system. The volumetric data can be obtained by repeated C-arm rotation with conventional 2D image acquisition, or by 270° rotation of the X-ray source and flat-panel detector in a continuous mode. For therapy guidance imaging, data acquisition time cannot exceed the time required for monitoring anatomic changes; hence, data are acquired in a single brief exposure. The most common approach uses a single rotation of the C-arm with an X-ray beam projecting a cone-shaped volume. Planar images are reconstructed with the pixel size of the image receptor or a larger pixel size for improved signal-to-noise ratio. Volumetric views of the region of interest are generated from the full volumetric dataset; 3D-CT reconstruction algorithms are embedded in dedicated commercial diagnostic workstations [235, 236, 237].

Surface imaging systems

Serving as non-invasive 3D optical scanners, play a crucial role in investigating patient setup variations during radiation therapy delivery. These systems gather 3D surface data, capturing the external geometry of the patient using visible light. Typically deployed in conjunction with CT imaging, the systems aid in transferring patient imaging information onto treatment machines.

The primary objective of patient surface imaging systems lies in the non-invasive and markerless identification of patient setup changes. Enhancements in the accessibility of optical devices have led to a rise in their application in clinical environments. Clinical workflows generally involve the acquisition of a volumetric CT image to delineate internal structures. Surface imaging is then performed with the addition of high-contrast surface markers, enabling the optical reconstruction of the external geometry. Deviation between the detected external surface and the CT-derived geometry—commonly the isocenter position within the volumetric image—is subsequently monitored. Deformations of the external shape are identified, often including non-rigid body transformations. The information can be integrated into treatment logs, guiding adaptive radiotherapy strategies.

Additional information can be derived by exploiting intrinsic textural features present on the patient. Depth-of-field limitations imply a blurred outer layer of the rendered surface geometry. This outer region can be masked, and a range image with the complete geometry can be generated. Information on the underlying anatomy sometimes contributes to the accurate prediction of the actual surface [238, 239, 240].

Adaptive radiotherapy

Enables patient-specific re-planning during treatment delivery to address anatomical and physiological changes affecting dose distribution. Imaging advancements allow for quality assurance and monitoring of treatment delivery. Treatment adaptation involves assessment of anatomical variation (e.g., weight loss) and functional imaging (e.g., PET) to customize dose distribution. An adjustment may occur before a new treatment course (e.g., head and neck) or at a lower frequency (e.g., daily - volume-of-the-day concept). Redesigned approvals and workflows incorporate these adaptive processes.

Significant disparities between treatment planning and delivery are viewed through an adaptive radiotherapy lens. Across anatomical sites, kV and MV imaging assist verification of bone structures. Cone-beam CT incorporates soft-tissue visualization and setup correction. Surface imaging system design reduces ionizing exposure quantified on a body-region · daily basis. These imaging capabilities imply within-day adaptation, integrating kV and MV images with known structures for correction strategies depending on tumor location and clinician preferences [241, 242, 243].

Chapter - 19

Quality Assurance in Diagnostic and Therapeutic Systems

Medical physics encompasses the sciences pertaining to radiation and matter interactions, physiological dosimetry, medical optical technologies, and medical imaging. It serves two main roles. In diagnostics, it applies physics to safeguard patients and staff during radiological procedures and optimize medical radiation dosimetry parameters. In radiation therapy, it covers management, operation, and quality assurance of systems and methodologies used to administer radiation for curative or palliative cancer treatments. A medical physicist applies physics in a clinical environment rather than pure academic or commercial research.

Sections of the International Commission on Radiological Protection (ICRP) and International Atomic Energy Agency (IAEA) reports detail principal missions, responsibilities, and functions of medical physics practice. The ICRP encompasses the establishment of the International System for Radiological Protection, all aspects of protection in radiotherapy and diagnostic x-ray procedures, and a recommendation on the protection of patients undergoing nuclear medicine procedures. The IAEA provides general guidance for the provision of clinical medical physics services at hospitals and surgery centres, accreditation and training requirements for clinical medical physics services, and guidelines for certification and habilitation of clinical medical physicists, radiation protection experts, and expert advisers in radiation protection. These reports outline the role, competence, and responsibilities of clinical medical physicists involved in medical radiation dosimetry, safety auditing, radiation therapy radiobiology, radiological image formation, and implementation of the principles of structured medical physics quality assurance [244, 245, 246].

QA for X-ray and CT systems

Conventions for X-ray, CT, MRI, ultrasound, and radiotherapy machine QA differ across institutions and may be specified by local health authorities. The principles outlined in AAPM TG 18 summarize recommended QA procedures for X-ray imaging systems. Quality assurance programs should include an evaluation of image quality at least annually, together with a detailed assessment of system performance whenever a major component is modified or replaced. Reference values and tolerance limits are given to enable comparison with baseline data accumulated over the operating life of the system.

The QA protocol is subdivided into five areas: (1) general machine and system performance tests; (2) image quality checks; (3) artefact detection; (4) indicator monitoring; and (5) post-processing evaluation. General machine and system performance tests cover: X-ray tube output (kVp, HVL); image receptor speed; impulse response; system stability; patient dose; dose linearity with input quantity; output with variable kVp; system compatibility with contrast media; and system transfer characteristics. Image quality checks include: image linearity; radiographic contrast performance; perceptual contrast; image resolution; stray radiation and field congruence; in-focus and out-of-focus detail definition; contrast-detail performance; and sugar-water or charcoal resolution. Image artefact detection assesses the presence of: grid use artefacts; scatter radiation; darkroom processing systems; and field alignment. Indicator monitoring covers: beam centring indicators; and preset kVp indicators. Post-processing evaluation examines the effect of post-processing algorithms on image quality and the potential introduction of unwanted artefacts.

For CT systems, AAPM TG 66 and the IAEA Technical Reports Series No. 457 describe a QA philosophy based on risk assessment principles. High-risk areas or categories that have been identified require routine monitoring and testing. Acceptance and QA tests should be performed at installation and any significant modification to the system; tests should also be conducted whenever there is a change of major hardware or software components. A test is also warranted when patient touring factors indicate the need [247, 248, 249].

MRI and ultrasound QA

Quality assurance (QA) for MRI systems encompasses visual inspections, field measurements, and quantitative tests. Safety procedures reduce the risk of injury to patients and operators. Standards are established or recommended by bodies such as the American College of Radiology (ACR), American Institute of Imaging Physics (AIIP), British Institute of Radiology (BIR), Canadian Association of Radiology (CAR), European Association of Nuclear Medicine (EANM), European Conference of Radiological Protection (ECRP), European Commission (EC), Institute of Electrical and Electronics Engineers (IEEE), International Atomic Energy Agency (IAEA), International Commission on Radiological Protection (ICRP), International Society for Magnetic Resonance in Medicine (ISMRM), National Electrical Manufacturers Association (NEMA), National Health Service (NHS), Royal College of Radiologists (RCR), and Royal College of Physicians and Surgeons of Canada (RCPSC).

The QA components generally include the Static Magnetic Field Strength measurement; the Measurement and Mapping of the Main Magnetic Field homogeneity; Gradient System Performance Testing; Radiofrequency Field (B1-) Homogeneity and Specific Absorption Rate (SAR); and Geometric Distortion. Image quality control measures include assessment of radiofrequency coils, assessment of contrast resolution, signal-to-noise ratio, and near-surface detection capabilities; quantification of geometric distortion; measurement of slice thickness; modulation transfer function; assessment of long and short T1 and T2 contrasts; and in-homogeneity artefacts. Quality assurance for ultrasound imaging systems comprises visual checks, testing of key parameters, assessment of image quality, and continuous monitoring of performance [250, 251, 252].

Radiotherapy machine QA

Protocols for regular quality assurance of radiotherapy machines are outlined in the American Association of Physicists in Medicine (AAPM) Report 100 (2009) and IAEA TRS 398 (2000). The AAPM report provides comprehensive checklists and acceptable tolerances for all radiotherapy machine types, in-vivo dosimetry checks, and clinical application considerations, while IAEA TRS 398 focuses on commissioning, reference dosimetry, output factors, and brachytherapy.

Quality assurance of X-ray and CT systems follows widely accepted international and national protocols, such as AAPM TG-18, AAPM Report 93, and IEC 61519 (2000). The protocols encompass visual inspections, performance tests with dedicated phantoms, monitoring of displayed quantities and reference signals, response to common modes of failure, and calibration. Many aspects of these protocols are also applicable to ultrasound and MRI systems [253, 254, 255].

Safety audits and accreditation standards

Safety audits and regulatory frameworks ensure that services journeying towards accreditation pay attention to patient, operator, and environmental safety. Facilities undertaking clinical procedures using X-ray and CT systems are expected to have a measurement system in place and follow the recommendations of the ICRP and ICRU on risk assessment. The planning phase of any clinical facility requires patient and operator safety to be paramount by ensuring that the radiation dose delivered is as lower as possible while achieving diagnostic requirements; the ALARA principle.

Following commissioning and before the commencement of clinical use, a safety audit is performed to ensure that potential hazards are being monitored and controlled. Quality assurance is an ongoing process with test protocols, manuals, and acceptance limits incorporated into all diagnostic and therapeutic procedures. Establishing accreditation standards for diagnostic radiology is the next logical step in the demand for quality health care; the end product of quality assurance is satisfied patients [256, 257, 258].

Chapter - 20

Future Trends in Medical Physics

Medical physics constitutes a diverse discipline involving diagnosis, therapy, radiotherapy, radioprotection, and the establishment of specialized techniques for the detection and correction of devices in those specialties. The goal is to guarantee patients, workers, and individuals exposed to medical practices adequate protection from radiation hazards without sacrificing image quality in radio-diagnostic applications or the success rate of radio-therapeutic procedures. While the basis of the work carried out by clinical medical physicists is common, each specialist typically dedicates their time and expertise to a specific area. Several references provide an overview of the function and activities of medical physicists in clinical practice, highlighting the regulatory, ethical, quality assurance, and quality control aspects.

Medicine has always been closely linked to the laws of physics, and it is logical that after establishing the fundamental principles of radio-activity during the late 19th and early 20th centuries the first practical applications would appear. These initiatives were essentially of a diagnostic nature, such as the photographic detection of a foreign body inside a hand or the use of the X-ray thallium method for diagnosing tuberculosis. Subsequently, with the production of therapy sources and units, the effort became even more impressive, with a great deal of work being done toward radiation treatment of different types of cancers. Despite the rapid evolution in diagnostic and therapeutic physics, the fundamental lessons learned were essentially empirical, arising from trial and error. Today, however, these practices have been refined substantially, and the fundamentals are well understood [259, 260, 261].

AI and machine learning in imaging and therapy

The application of artificial intelligence (AI) and machine learning

(ML) in medicine is rapidly expanding with potentially far-reaching consequences. Although labeled "artificial," the effects of any intelligent system—human or machine—ultimately rest on its interaction with humans. Like any other tool, AI and ML can be applied poorly, with little real understanding of its effects or limitations. The future usefulness of AI systems in clinical practice rests on the validation that specialists in the specific area require and their ability to produce better results than conventional methods of image processing, interpretation, and treatment planning. In image acquisition, use and detection of radiation dose, localization of tumors and lesions, prediction of patient outcome, and treatment planning, AI and ML will undoubtedly have a major impact on dosimetry and treatment planning, in developing new visualization techniques, and in identifying patients at greater risk of complications by combining different types of clinical data.

A second area where great hopes are being placed for higher efficacy is in the use of proton and heavy-ion therapy, which results in skin-sparing radiotherapy with a higher ion density at the tumor site. With protons and heavier atoms, a much higher LET is achieved, with consequent advantages and disadvantages for the irradiation and LQ models. The particles used have the same order of magnitude as cellular diameters, enabling greater ionization densities. Greater initial slowing down of heavy atoms and ions results in more secondary-atom production, which augments the radiobiological effect. In addition to proton therapy, hadron therapy (C and O competition at higher energies) and most recently RBE-modulated ion therapy are presented. Attention is also given to hybrid imaging, with the advantages offered by the combination of two types or more; for example, by associating the anatomical information from a CT scan with functional information from a PET scan, it will be possible to increase the accuracy of the definition of areas with different biological behavior. Finally, personalized approaches to radiotherapy and adaptive radiotherapy are examined, both concepts that adapt the treatment during the course of the therapy by taking into consideration different changes [262, 263, 264].

Proton and heavy-ion therapy

Modalities exploit the finite Bragg peak characteristic of particle beams to treat patients with tumours located in normal tissues, especially in paediatric and young adult cases. Heavy-ion radiotherapy has been adopted only in a few centres owing to the low availability of synchrotrons and cyclotrons as well as to the limitations of the technique in terms of intra- and post-therapy imaging, treatment control, and on-line adaptive processes that are becoming routine in X-ray therapy. While several advantages of proton therapy over classical X-ray therapy remain to be conclusively demonstrated, proton therapy centres continue to be built at a very fast pace. These facilities raise medical physics challenges in terms of the quality assurance of proton beam production, monitoring, and delivery.

Proton and heavy-ion therapy share with external X-ray therapy the necessity to account for patient geometry, tissue density, and electron density heterogeneities in the dose calculation process. Optimisation approaches such as 3D conformal radiotherapy, intensity modulated radiotherapy, or volumetric modulated arc therapy have been developed and are widely used for proton treatment planning. Such advancements have permitted the successful application of proton therapy approaches different from the classical spread-out Bragg peak such as non-homogeneous dose distributions, boost treatments, and combined X-ray and proton therapies [265, 266, 267].

Hybrid imaging technologies

Two modalities are said to be hybrid when they are combined into one system to exploit their strengths and overcome their weaknesses. The use of hybrid imaging techniques improves the quality of the images without adding new equipment, saving on cost and space [268]. For instance, one possibility is to combine two modalities that do not interfere with each other. One option is to measure the parameters provided by one modality after the measurements of the second one have all been acquired [269].

Personalized and adaptive radiotherapy

Enables timely access to treatment when patients are at greatest risk of cancer recurrence, improving safety and reducing treatment burden for precancerous lesions while considering the risk of new lesions. Tumors often exhibit growth between planning and treatment stages, which may necessitate replanning irrespective of radiotherapy technology advances. Adaptive radiotherapy uses information acquired during treatment, such as imaging showing tumor positional deviations, to modify the original plan to account for the detected changes [270].

Magnetic resonance imaging (MRI) guidance captures both initial tumor position and ongoing tumor motion without excessive radiation exposure. Adaptive radiotherapy based on MRI guidance facilitates proton and photon delivery techniques by adjusting isodose distributions after monitoring tumor shifts. The resulting safety increase while maintaining treatment effectiveness extends the use of fractionated schedules to tumors that would otherwise receive intra-fraction irradiation. Techniques such as respiratory motion management improve precision by quantifying tumor movement as major organs shift over time, and methods for real-time intrafraction re-planning and adaptive intensity-modulated radiotherapy (IMRT) sequencing are under development to further increase treatment speed and precision. Furthermore, MRI-guided radiation therapy permits intra- and inter-fraction plan adaptation without preliminary planning steps, widening the therapeutic window and addressing previously disallowed cases. The R-IDEAL framework supports systematic evaluation of emerging technologies and techniques across the spectrum of image guidance in radiotherapy, specifying areas such as imaging, tracking, characterization, and adaptation. In prostate treatment, MRI guidelines are identified as game changers enabling safer reirradiation by improving quality and effectiveness monitoring. MRI-based intensity histogram analysis predicts treatment response in rectal cancer, enriching personalized therapy planning [271, 272, 273].

Conclusion

Fundamentals of Medical Physics in Diagnostics and Radiation Therapy

Medical physics encompasses the application of physics to medicine across four major domains of health care: diagnostic imaging, radiation therapy for the treatment of cancer, radiation safety and protection, and the development of new medical technologies related to imaging, therapy, and sensors. Each of these domains has a substantial associated scientific literature and thus is studied separately. Their common theme is the application of the principles of physics to health care, while drawing on a foundation of biology, chemistry, physiology, and related disciplines.

The introduction of qualitative physics into the clinical sciences allows a better understanding of concepts, modalities, and methods that intimately rely on the knowledge of nature phenomena and their development. Quality assurance, regulatory requirements, accreditation systems, and the introduction of second opinions come later in the training process and should improve the safety and quality of medicine.

A series of important concepts related to ionizing radiation are fundamental to nursing and general basic medical education and namely atomic structure and the principles of interaction of photons with matter; dosimetry and dosimetry related quantities; X-ray production and tube physics; all stages of image formation in projection radiography; detector technologies in radiography and fluoroscopy; clinical applications of radiography and fluoroscopy; the basic principles of ultrasound physics and ultrasound imaging; basic principles of nuclear medicine and radiopharmaceuticals; biological effects of ionizing radiation; basic principles of external beam and brachytherapy radiation therapy treatments; therapy imaging; and finally quality assurance for medical equipment. Basic concepts in magnetic resonance imaging physics and emerging technologies such as virtual and augmented reality, artificial intelligence, or hybrid imaging and associated applications, are also relevant.

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