Principles of CT and MR imaging

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Learning Objectives

- To understand the physical principles of interaction between X-rays and tissue
- To learn the basic principles of image formation in X-ray computed tomography
- To learn the basic principle of radiation dose estimation and measurements
- To learn the basic principles of MRI: magnetic field, sequences, signal parameters, and coils
- To learn the safety risks in MRI

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8.1 Physical Principles of CT

The formation of CT images involves measurement of the X-ray transmission profile through a patient over a large number of views. Each profile is acquired by means of a fan-shaped X-ray beam that penetrates the body. The X-rays exiting the scanned patient's body are recorded by a detector arc, generally consisting of 800–1000 detector elements, referred to as a detector row. Each projection of detector's record (transmission profile) is a view at one position (angle) of the X-ray source. Multiple angular profiles are collected during one complete rotation of the X-ray tube around the patient, which are used to reconstruct the CT image (basically a square matrix of voxels) of the internal organs and tissues for each complete rotation of the X-ray source.

Along the pathway through the body, X-rays interact with the matter. In the energy range typically used in a CT scanner, there are two principal ways in which X-rays interact with tissues: photoelectric effect and Compton (incoherent) scatter. The image signal relies on the photoelectric effect, whereas Compton scatter degrades the image quality by adding noise.

The main effect of X-rays' interaction with matter is attenuation of the X-ray beam; thereby the detector records the mean absorption by all the different tissues which the X-ray beam has penetrated. "Computed tomography, on the other hand, measures the attenuation of X-ray beams passing through sections of the body from hundreds of different angles, and then, from the evidence of these measurements, a computer is able to reconstruct pictures of the body's interior." [1]

In the reconstructed image, each voxel is associated with the linear attenuation coefficient μ (m^{-1}) of the corresponding tissue. The linear attenuation coefficient μ is a function of photon energy and material density. The X-ray attenuation through the body is governed by the Beer-Lambert law.

8.1.1 Hounsfield Unit

Since the earliest days of CT, the attenuation value calculated for each voxel of the reconstructed matrix is replaced with an integer number called Hounsfield unit, in honor of its inventor. Large part of the human body is made up of water and surrounded by air, and this was the biological background for creating the linear Hounsfield scale. The Hounsfield unit (HU) scale is expressed relative to the linear attenuation coefficient of water at room temperature.

Distilled water at standard pressure and temperature is defined as 0 HU; the radiodensity of air is defined as −1000 HU. Density of other tissues is related to this range, usually from −1000 HU to +1000 HU. HU = 1 is associated with 0.1% of the linear attenuation coefficient of water. All substances except water and air might show variations in HU values when they are evaluated at different tube voltages. Usually, for each tube voltage, the CT vendors carry out an adequate calibration of the HU scale by acquiring images of air- and water-filled phantoms.

Key Learning Points

- The image signal relies on the photoelectric effect, whereas Compton scatter degrades the image quality by adding noise.
- Computed tomography measures the attenuation of X-ray beams passing through sections of the body from hundreds of different angles.
- The Hounsfield unit (HU) scale is expressed relative to the linear attenuation coefficient of water at room temperature.

8.2 The CT Imaging System

8.2.1 Current Acquisition Configuration

Since its first appearance in the 1970s, CT systems developed rapidly as a valuable tool in diagnostic radiology. The geometry of modern CT systems is still similar to scanners sold in late 1975: a full X-ray fan beam that covers the complete field of view and a simultaneously rotating tube and rotating detector array [2]. In 1987, continuously rotating systems based on slip-ring technology became available. Typical rotation time was 1 s. Helical CT was introduced in 1989 and led to a breakthrough in truly three-dimensional (3D) CT imaging. Following the development of larger detector arrays with more detector rows, dual-source CT (DSCT) was introduced in 2005. Such technology was intended to further increase scan speed and temporal resolution, two requirements for cardiac CT.

To date the CT source and detection technologies have evolved to satisfy three major CT imaging concerns: (1) increasing the axial field of view, (2) decreasing the rotation time to improve temporal resolution and scan speed, and (3) last but not least, reducing the radiation dose to the patients. The emergence of multislice CT scanners was possible, thanks to solid-state detectors segmented into detector elements arrays. CT sources evolved to support larger coverage per rotation. The high speed of acquisition required X-ray sources capable of supporting great accelerations and increased peak power, while at the same time, the detection systems were designed to have short integration periods. Furthermore, over the last few years, the availability of iterative reconstruction made the use of low and ultra-low dose acquisitions possible to be used routinely in clinical practice. This has a direct impact on the need for lower noise detection systems [3].

8.2.2 Gantry Technology

The introduction of slip-ring technology to transport data and energy to and from the gantry enabled continuous rotation of the X-ray tube and detector. This was an essential prerequisite for the development of spiral CT scanners. Slip rings are electromechanical devices made up of circular electrical conductive rings concentric to the gantry axis and brushes that transmit electrical power and all of the control signals from the stationary parts of the gantry to the rotating frame. These technological improvements allow the scan frame to rotate continuously with no need to stop between rotations to rewind the system cables [4].

The latest CT systems with gantry rotation times down to 0.25 s use noncontact data and energy transfer between the rotating and stationary parts of the gantry. Rotating hardware undergoes huge centrifugal forces on the order of 40 × *g*. This significantly affects the X-ray tube design for mechanical reasons. Larger patient body sizes and the need to perform whole-body scans additionally require new innovations regarding the patient table. To realize fast and particularly high-pitch scanning (table speeds of up to 74 cm/s are currently in use), the acceleration and deceleration range of the table must guarantee patient safety and comfort.

The CT table has also been adapted to handle increasing patient's body weight. Larger patients require expanded table capacity limits as well as more noteworthy gantry width (up to 90 cm in dedicated systems), broadened reconstruction field of view, and higher tube output. Also, a larger bore width provides more space to CT-guided interventions as well as uncompromised patient positioning [2].

8.2.3 Tube Technology

X-ray tubes are subjected to far higher thermal loads in CT than in any other diagnostic X-ray application. Shorter scan times in current CT scanners require high-power X-ray tubes and utilization of oil-cooled rotating anodes for efficient thermal dissipation. Typical maximum tube power is on the order of 100 kW or more. The advent of helical CT with nonstop scanner rotation has introduced additional requirements for X-ray tubes.

Several technical advances in component design have been made to achieve these power levels and manage the issues of target temperature and heat dissipation. As scan times have diminished, the anode heat capacities have expanded by as much as a factor of 5, avoiding the need for cooling delays during most clinical procedures. Tubes with capacities of 5–8 million heat units are available. Furthermore, improvement in the heat dissipation rate (kilo–heat units per minute) has increased the heat storage capacity of modern X-ray tubes. The huge heat capacities are achieved with thick graphite backing of target disks, anode diameters of 200 mm or more, improved high-temperature rotor bearings, and metal housings with ceramic insulators among other factors. The working life of current tubes ranges from 10,000 to 40,000 h, compared with the 1000 h typical of conventional CT tubes [5].

8.2.4 Detector Technology

The CT detector converts the X-rays into visible light, which is in turn converted into an electric current. Typically a CT detector is made up of three essential layers: (1) a scintillator to convert X-rays into light, (2) a photodiode to convert light into current, and (3) a substrate to provide electrical and mechanical infrastructure.

8.2.5 Scintillators for CT

A scintillator is a luminescent material able to convert linearly high-energy photons into visible light. The scintillator is optically coupled to matching silicon p–i–n photodiode matrices. The requirements for the scintillator material to be used in X-ray CT can be summarized as follows [6]:

- High absorption for X-rays in the energy range up to 150 keV
- High light output
- Radioluminescence spectrum in the visible or near IR range to match the spectral sensitivity of silicon photodetectors
- Decay time in the range of 1–10 μs
- No afterglow or ghosting
- Good radiation hardness for high X-ray fluence
- Small temperature dependence of the light yield
- Good mechanical properties
- Affordable cost

These requirements make polycrystalline ceramics and single crystals the most convenient types of scintillators. Scintillators for multislice CT scanner are built in two-dimensional (2D) arrays, with a pixel size around 1 mm. The arrays packaging also incorporates a reflective material. The reason for the reflective material is both to mechanically support the pixelated scintillator array and to efficiently transport the scintillation light to the photodetector, with minimal cross talk [3].

8.2.6 Photovoltaic Detector Array (PDA)

To date the light exiting from scintillators is collected by a photovoltaic detector array (PDA), which converts it into an electric current. In traditional detector design, the small analogic electrical current from the photodiode was amplified and changed into a digital signal on an external board, requiring analogic connections between the detector and the electronic circuit components on the external board. These types of detectors are referred to as conventional detectors. Current commercial CT systems are equipped with detectors featuring

fully integrated electronics. In these detectors, the photodiode, the preamplifiers, and the analog-to-digital converters are integrated into the same silicon chip that is attached to the scintillation ceramic, thus avoiding the need for analog connections. Such integration reduces the time during which the signal is in an analog format, so the amount of electronic noise that can be superimposed to the signal is reduced. These types of detectors are referred to as integrated detectors [7].

Currently, semiconductor-based direct converting detectors are under development for CT imaging. With such as innovation, each detected X-ray photon generates a very short pulse enough to count each photon. In addition, the area under the pulse, and thus the pulse height, is proportional to the photon energy. The goal is to design photon-counting detectors with energy discrimination capabilities to capture spectral X-ray information without the need to apply two different X-ray tube voltages [2].

8.2.7 Anti-Scatter Grids

Scatter is a significant source of image artifacts in CT. Scatter management includes both scatter rejection and scatter correction via software. Scatter reduction becomes even more important for large CT detectors with more than 64 rows, because image quality is influenced by the increasing quantity of scattered photons for large cone angles. Furthermore, larger patients increase the scatter problem that degrades image quality.

Conventional CT scanners (with detector coverage not exceeding 40 mm along the patient axis) usually employ one-dimensional (1D) anti-scatter grids (ASG) mounted in front of the detector. Scattered X-ray photons are suppressed by using lamellas made of X-ray absorbing material positioned between the active detector cells [8]. Materials used to make anti-scatter lamellas are of high Z numbers in order to allow effective absorption of scattered radiation. To ensure adequate image quality in wide-cone CT, a two-dimensional (2D) ASG is needed [9].

Key Learning Points

- Geometry of modern CT systems is based on a full X-ray fan beam that covers the complete field of view, coupled with a simultaneously rotating tube.
- Current CT technology is designed to increase the axial field of view, shorten the rotation time (to increase time resolution and scan speed), and to reduce the radiation dose to the patients.
- A combination of high-speed gantry rotation, tubes with high heat dissipation rate, high performance scintillators, and new detectors with fully integrated electronics, are the key components of modern multirow CTs.

8.2.8 Multi Detector CT

The term multislice CT (MSCT) scanner refers to a CT system in which multiple-row detector arrays are assembled to simultaneously acquire data at different slice locations [10]. Clinical applications benefit from multi-detector row CT (MDCT) technology in several ways: (1) shorter scan time, (2) large longitudinal (z) scan, (3) continuous data acquisition (enabling the images to be reconstruct at any z position), and (4) improved longitudinal resolution.

The CT detectors must provide different slice widths to adjust the optimum scan speed, longitudinal resolution, and image noise for each application. Earlier CT scanners were equipped with a single detector. Different slice widths were acquired by means of pre-patient X-ray collimation. To acquire more than one slice per rotation, CT scanners were then designed with a greater number of detectors. In 1998, all major CT vendors introduced MSCT systems, which typically offered simultaneous acquisition of four slices at a rotation time of down to 0.5 s [11]. Since then, the evolution of CT acquisition has followed its own form of "Moore's Law"; in particular, growth in the number of slices has been exponential, doubling every 18 months. As a next step, the introduction of an 8-slice CT system in 2000 enabled shorter scan times but did not yet provide improved longitudinal resolution (thinnest collimation 8 × 1.25 mm). The latter was achieved with the development of 16-slice CT, which made it possible to routinely acquire substantial anatomic volumes with isotropic submillimeter spatial resolution [12]. In 2004 the next generation of multislice CT systems became commercially available, with 32, 40, and even 64 simultaneously acquired slices, with a strong impact on volume coverage speed. Similar to some established 4-slice and 16-slice CT systems, the 64-slice scanner has an adaptive array detector design with different sizes of the detector rows in the longitudinal direction. The innermost elements can be used to collect thin slices or combined in pairs to acquire thicker slices. Some vendors make use of a periodic motion of the focal spot in longitudinal direction (z-flying focal spot) to improve data sampling along the z-axis and to double the number of simultaneously acquired slices. To date the 320 detector row CT scanner represents the current pinnacle in the nominal number of slices that can be scanned per gantry rotation. A detailed story of the race for the increase of the axial coverage in clinical CT scanners can be found in [13].

After 2007, the clinical market has reached a plateau in terms of number of simultaneously acquired slices, considering some issues about patient dose, image quality, and cost. On the other hand, the advent of dual-energy CT (DECT) seems now to have pulled manufacturers into a new battlefield: the number of energy bin must grow instead of the number of slices, even though true energy-resolved scanners based on photon-counting detectors are not yet mature for clinical use.

8.2.9 Adjustable Parameters for CT Image Acquisition

The common CT acquisition parameters are:

- *Tube current-time product* (milliampere second, or mAs) is the product of the X-ray tube current (in milliamperes) and the CT scanner exposure time per rotation (in seconds).
- *Tube current modulation* (mA modulation) is an essential tool to ensure proper patient exposure during CT examinations. It allows the tube current to be actively modulated during the scan to reduce radiation to the patient.
- *Tube voltage* (kV): The X-ray tube potential indicates the peak energy of the X-ray photons (in kilovolts) in a spectrum of X-ray energies.
- *Tube voltage modification* (kV modification): Different tube voltages can be selected for the CT examination depending on the size of the patient or the type of CT examination being performed. Lowering tube voltage from 120 to 100 or 80 kV makes it possible to reduce the radiation dose provided to the patient and is advised for small and average-sized patients.
- *Effective tube current-time* product (effective mAs) is described as the ratio of tube current-time product to pitch. With this concept, the variation of pitch is exactly compensated by changing tube current or rotation time to maintain the same image quality [14].

8.2.10 Pitch

Helical scanning with multislice CT scanners is indistinguishable from that with single-slice CT scanners: table movement and tube-detector rotation occur simultaneously with continuous data acquisition.

For single-slice CT, detector pitch was defined as table movement per rotation divided by slice thickness. For instance, with a slice thickness of 10 mm and a table movement of 15 mm per rotation, pitch would be 1.5.

In multislice CT, beam pitch is defined as table distance traveled in one rotation divided by total thickness of all simultaneously acquired slices. Pitches of greater than 1 mean that there are gaps between the X-ray beams from adjacent rotations. Pitches of less than 1 mean X-ray beam overlap (and thus double irradiation of some tissue) and so are generally not clinically used [15].

Key Learning Points

- The term multislice CT (MSCT) scanner refers to a CT system in which multiple-row detector arrays are assembled to simultaneously acquire data at different slice locations.
- The CT acquisition parameters are: Tube current-time product, CT scanner exposure time per rotation, Tube current modulation, Tube voltage, Tube voltage modification, Effective tube current-time product.
- The pitch is the table distance traveled in one rotation divided by total thickness of all simultaneously acquired slices.

8.2.11 Image Reconstruction

Image reconstruction in CT is a mathematical process that generates tomographic images from X-ray projection data acquired at many different angles around the patient.

Analytical reconstruction algorithms, among these the Filtered Back Projection (FBP), assume that both the projection data and the measurement process can be described by continuous functions. In a simplified model, a projection is obtained by drawing a set of narrow (pencil beam) X-rays through the patient. Each single pixel in the projection is the integral of the patient's density along each pencil beam. A single projection of the patient is one-dimensional. CT reconstruction of the patient requires several projections at different angles between the pencil beam and the patient. The X-ray source is rotated by an angle α and a new projection is acquired, and the process keeps going until all the necessary projections are acquired.

The collection of projections at several angles is called sinogram. The inverse Radon transform is used in CT to reconstruct a 2D image from the sinogram. The FBP is the fastest tool to perform the inverse Radon transform. Each projection is convoluted to a filter to compensate for the effect of the so-called low-pass blur that occurs because of the different numbers of projections passing through the center and the periphery, respectively, of an object.

In clinical practice, several filters (kernels) are available to meet the requirements of spatial resolution and image noise. Increasing the strength of the low-pass blur increments the "sharpness" of the image, as well as expands image noise. Different kernels allow optimized depiction of high-contrast structures or soft tissue, such as lung tissue or bone. Image sharpness and image noise are intrinsically coupled in image reconstruction via FBP: the sharper the image, the higher the image noise. A major limiting feature of FBP is that it misses the capability to account for image noise resulting from Poisson statistical variations in photon number over the image plane and system hardware details (focal spot size, active detector area, and image voxel shape); in practice, the effort to reduce the radiation dose to patients translates into an increase in image noise. Delineation and low-contrast detectability of a structure suffer from high image noise and artifacts, so that to generate a diagnostic CT data set the radiation dose may not be lowered below certain minimal constraints. Lowering image noise by choosing "smoother" kernels for image reconstruction will result in impaired spatial resolution when using a conventional FBP technique. [16]

The recent augmented computational capacities in normal workstations and the on-going attempts to reduce doses in CT have focused the attention of manufacturers on iterative reconstruction (IR). All IR methods consist of three steps which are repeated iteratively. First, a forward projection of the object estimate creates synthetic raw data which, in a second step, are compared to the real acquired raw data in order to calculate a correction term. In the last step, the correction term is back-projected onto the volumetric object estimate. The iteration process can start with an empty image estimate or using prior information, for example, a standard FBP reconstruction or a volume of a similar object. In general, the process may converge faster toward a stable solution if the initial images match the final images. The iterative process ends when either a maximum number of iterations are reached, or the update for the current image estimate is considered small enough, or when a predefined quality criterion in the image estimate is satisfied. More detailed information about IR may be found in [17].

Iterative reconstruction algorithms have several advantages compared to analytical FBP-based methods: different physical models might be integrated in the reconstruction process, thus reducing image noise and artifacts depending on the degree of modelling. They do not introduce new artifacts related to approximations that are often exploited in analytical reconstruction methods, and they are more appropriate for dealing with missing data or irregular sampling. The biggest advantage of IR methods is generally considered the intrinsic potential for patient dose reduction.

All iterative reconstruction methods have the main disadvantage of requiring increased computational effort that becomes necessary due to multiple iterations instead of a single iteration as for analytical methods.

Also the image quality of reconstructed image depends on the prior information used for the parameterization of the acquisition model, and the necessary number of iterations might vary strongly depending on the measured object. Different noise patterns and some artifacts that manifest themselves in a way different from analytical reconstruction methods had impact on the acceptance in clinical routine of images reconstructed by IR methods.

8.3 CT Dose Index

8.3.1 Dose in CT, CTDI, $CTDI_w$, DLP, and SSDE

Interest in radiation exposure during CT examinations has increased during the last years. Scientific literature has recently highlighted the importance of radiation dose from X-ray CT examinations and the associated risk. The dose level imparted in CT is much greater than those from conventional radiology and the use of CT keeps growing, often by 10–15% per year. Thus, among medical procedures involving ionizing radiation, CT will be responsible for the major portion of the total collective dose delivered to the public [18].

CT manufacturers designed the CT scanner to keep the radiation dose to the patient as low as reasonably achievable (ALARA). X-ray source, detectors, and reconstruction algorithm are involved in the process of dose reduction. Following CT evolution over the years, physicists developed new dose indexes to evaluate and communicate the dose imparted in CT examinations. The AAPM report 96 and the AAPM report 204 are the most valuable references for detailed and standard definition of CT dose index currently in use. Report 96 also provides an overview on the methods for dose reduction in CT. Hereinafter, the most commonly used CT dose index parameters are discussed.

The primary quantity for assessing the dose in CT is the computed tomography dose index (CTDI). CTDI represents the average absorbed dose, along the z-axis, from a series of contiguous irradiations. It is measured from one axial CT scan and is calculated by dividing the integrated absorbed dose by the nominal total beam collimation.

$CTDI_{100}$ is a linear measure of dose distribution over a 100 mm length pencil ionization chamber; since this parameter underestimates dose for longer scan lengths and does not take into account the topographical variations of a human body, it is not in clinical use. In fact, the $CTDI_{100}$ across the field of view (FOV) is not constant. For instance, for a body CT imaging, the $CTDI_{100}$ at the surface is almost double or higher than the $CTDI_{100}$ at the center of FOV. The weighted $CTDI_w$ is defined as the sum of one-third the $CTDI_{100}$ at the center and two-thirds the $CTDI_{100}$ at the edge of a phantom. $CTDI_w$ is closer to the human dose profile as compared with the $CTDI_{100}$.

To represent dose for a specific scan protocol, which almost always involves more than one scan, it is essential to consider any overlaps or gaps between the X-ray beams from consecutive rotations of the X-ray source. By dividing the $CTDI_w$ by the pitch factor, one obtains the $CTDI_{vol}$. While $CTDI_w$ is representative of the average absorbed dose over the x and y directions at the center of scan, $CTDI_{vol}$ is the average absorbed radiation dose over the x, y, and z directions. This index was designed to offer a standardized method to compare radiation output levels between different CT scanners using a reference phantom, usually a 16 cm and 32 cm diameter polymethyl methacrylate (PMMA) cylindrical phantom often called “head” and “body” CTDI phantom, respectively. The SI units are milligray (mGy).

To better represent the overall energy imparted by a given scan protocol, the absorbed dose can be integrated along the scan length to compute the dose length product (DLP).

DLP is related to the total ionizing energy delivered to the reference phantom attributable to the complete scan acquisition and depends on all the adjustable acquisition parameters (kVp, mAs, pitch, slice thickness, and so on).

The $CTDI_{vol}$ or DLP, as visualized on consoles, do not represent the actual dose administered to the patient; they should be considered as an index of radiation output by the system, which is useful only for comparison purposes.

Given the huge variability in patient size, $CTDI_{vol}$ is corrected for the patient size, thus yielding the size-specific dose estimate (SSDE); more information about SSDE may be found in AAPM report 204.

It is important to notice that the potential biological effects from radiation depend not only on the radiation dose to an organ or tissue, but also on the biological radiosensitivity of the tissue or organ irradiated. Effective dose, *E*, is the dose descriptor that reflects this difference in biologic radiosensitivity. The units of effective dose are sieverts; generally millisieverts (mSv) are used in diagnostic radiology. "Effective dose aims to provide a single number that is proportional to the radiobiological detriment from a particular, often inhomogeneous, type of radiation exposure" [19].

Key Learning Points

- In CT images are reconstructed through a mathematical process that generates tomographic images from X-ray projection data acquired at many different angles around the patient.
- The primary quantity for assessing the radiation dose in CT is the computed tomography dose index.
- The absorbed dose can be integrated along the scan length to compute the dose length product (DLP).

8.4 Physical Principles of MRI

8.4.1 The Source of the MR Signal

During the first half of the twentieth century, Rabi, Bloch, and Purcell discovered the physical principles of magnetic resonance imaging (MRI). MRI is now an essential imaging tool in modern medicine to image anatomy and other functions within the body. MRI is capable of producing images of a patient exploiting the abundance of hydrogen atoms within the tissues, mainly from water and fat. In contrast to X-ray computed tomography, MRI is based on radio-frequency electromagnetic fields (RF-EMF). Typically, RF-EMF do not have enough energy to ionize the matter, thus avoiding the risk of X-irradiation. Quantum mechanics describes what happens to tissues when they are placed in a magnetic field. A simple classical model may be sufficient to explain all the theory behind MRI. According to classical mechanics, an object rotating around its axis carries a quantity known as *angular momentum* that can be thought as a rotational inertia, reflecting the object's size, shape, mass, and rotational velocity. It is typically represented as a vector pointing along the axis of rotation.

Protons, neutrons, electrons, and whole nuclei show an analogue property known as spin, usually visualized as a small sphere that rotates at a high speed around its axis even if such a picture is not strictly correct. Spinning masses generate an angular momentum. Since the single proton of hydrogen carries a positive charge, its spin generates an electrical current similarly as a charge in a looped wire. This loop current provides the proton a torque, referred to as *magnetic momentum*, when placed within a magnetic field [20].

Nuclei useful for MRI must have both magnetic moment and angular momentum; typically this happens if nuclei have odd-numbered atomic mass: ^{1}H, ^{13}C, and ^{31}P. Such nuclei are said to have nuclear magnetic resonance (NMR) property and are generally referred to as spins. In the simplest classical model nuclei having NMR property may be imagined as small magnets or dipoles. It is well known that if a compass is placed close to a powerful magnet, the compass needle aligns itself with the field in any orientation. If an external magnetic field ($\mathbf{B}_0$) is applied, the hydrogen proton having a spin quantum number ±½ tends to align with the field, but, as known from quantum theory, only two states are allowed: with and against the direction of $\mathbf{B}_0$. In particular, the angle between the spin and the magnetic field $\mathbf{B}_0$ is 54° 44′ [21].

Each state corresponds to an energy level: spins pointing with $\mathbf{B}_0$ (also referred to as parallel or spin-up) are in a lower energy level, which is the preferred alignment; spins pointing against $\mathbf{B}_0$ (also referred to as antiparallel or spin-down) are in a higher energy level. The energy-level difference, or energy gap, is given by the product of the proton gyromagnetic ratio, the external magnetic field intensity $|\mathbf{B}_0|$, and the reduced Planck constant $\hbar$. Once known the energy-level difference, which depends on the magnetic field strength, the distribution at room temperature of spin-up and spin-down may be calculated by means of Boltzmann equation. For commonly used magnets, operating at 1.5 T (Tesla), the population ratio is approximately 1,000,000–1,000,009 [22]. Thus, at 1.5 T, only nine hydrogen protons out of 1,000,000 are oriented with $\mathbf{B}_0$. That is why NMR is a fairly insensitive modality. Considering the huge number of hydrogen nuclei (about 10^{27}) available in the body, the net magnetization is still measurable. It is proportional to the field strength: a large field produces a greater magnetization and better signal-to-noise ratio [23]. This may explain why image quality increases at higher field strength.

Averaging the orientations of these few spins makes the net magnetization vector $\mathbf{M}$ to appear; this vector can have any orientation with respect the magnetic field. The net magnetization vector has a wobbling or precessional motion around $\mathbf{B}_0$ at the Larmor frequency.

The central equation of MRI $\omega = \gamma|\mathbf{B}_0|$ is known as Larmor equation, where γ is the gyromagnetic ratio and $|\mathbf{B}_0|$ is the intensity of external field. The frequency ω is also known as *resonant frequency* or *Larmor frequency*.

Transitions between the spin-up and the spin-down state can be induced by the absorption or the emission of electromagnetic radiation of angular frequency exactly equating

the Larmor frequency. When the magnetic field is a few Tesla, the Larmor frequency may range between 1 and 100 MHz. More specifically, at 1 T the Larmor frequency is 42.6 MHz [24].

Key Learning Points

- MRI produces images of a patient exploiting the abundance of hydrogen atoms within tissues, mainly from water and fat.
- Nuclei useful for MRI must have both magnetic moment and angular momentum; typically this happens if nuclei have odd-numbered atomic mass, such as ^{1}H, ^{13}C, and ^{31}P.
- The fundamental equation of MRI $\omega = \gamma|\mathbf{B}_0|$ is known as Larmor equation, where γ is the gyromagnetic ratio and $|\mathbf{B}_0|$ is the intensity of the external field.
- Frequency ω is also known as *resonant frequency* or *Larmor frequency*.

8.4.2 Signal Formation

Seen from a frame of reference rotating at the common frequency, the net magnetization vector **M** arising from the hydrogen nuclei inside the magnetic field in its equilibrium state is stationary, "so even though the individual spins are precessing, there is no net emission of signal in equilibrium" [25]. Thus, to obtain useful information from the precessing protons, we need to alter the direction of **M**. Radio-frequency (RF) energy pulses of exactly the Larmor frequency are used to modify the **M** direction. [26]

RF energy pulses ($\mathbf{B}_1$) produce two main effects on the protons inside a magnetic field: a certain portion of protons are inverted into the excited state spin-down (oriented against $\mathbf{B}_0$), thus giving rise to a net magnetization vector oriented in the plane transverse to $\mathbf{B}_0$. Phase coherence is established as well; this means that nuclear spins are precessing synchronously. If the $\mathbf{B}_1$ field is continuously applied, the net magnetization **M** is pulled away from its initial alignment with $\mathbf{B}_0$. **M** can be rotated 10°, 90°, 180°, 270°, or any amount. After every 360° rotation, **M** returns to its initial alignment with $\mathbf{B}_0$. The rotating transverse magnetization can generate a signal in the receiver coil around the patient or close to the patient's surface.

Soon after the RF $\mathbf{B}_1$ is turned off, the net magnetization **M** tends to realign itself with the external magnetic field $\mathbf{B}_0$ to recover the thermal equilibrium. This means that the magnetization decays over time, which implies that the magnitude of rotating transverse magnetization decreases, thus the signal induced in the receiver coil around the patient will diminish over time. The phenomenon is referred to as free induction decay (FID). The time necessary to recover the thermal equilibrium is the relaxation time, which is constant for a given substance at a given magnetic field strength.

Two independent relaxation processes exist: longitudinal relaxation and transverse relaxation. Both processes are characterized by a particular time. The time taken by **M** to recover 63% of its equilibrium value after a 90° rotation is called T_1, or longitudinal relaxation time, and arises from the interaction between the spins and the atomic neighborhood. The second relaxation process that occurs simultaneously to T_1 relaxation is due to the spin-spin interaction and inhomogeneity of the main static magnetic field $\mathbf{B}_0$. It affects the phase coherence with which the spins precess around the magnetization vector. Thus, as time passes, the signal recorded by the coil decreases because the spins begin to lose their phase coherence (i.e., one spin becomes out of phase with another spin).

Phase dispersion is mainly attributable to variation in the local precessional frequencies induced by random local magnetic inhomogeneities. Transverse relaxation is characterized by T_2 relaxation time. T_2 relaxation time is the time at which the signal decays to 37% of its original value. T_2 time is always shorter than T_1 time [26]. In practice, the signal decays with a faster time constant T_2 because $\mathbf{B}_0$ is not uniform.

8.4.3 Signal Localization and Image Formation

Gradients are employed to localize the MR signal in a region of interest. Gradients are additional linear variations of the magnetic static field strength in a selected region. Gradients can be applied in any orthogonal direction using the three sets of gradient coils, G_x, G_y, and G_z, within the MR system. In presence of gradients, the frequency of precession will be different at different positions in the patient. Faster or slower precession is detected as higher or lower MR signal. Thus, the frequency measurements can be used to distinguish MR signals at different positions in space and enable image reconstruction in three dimensions [27]. Image reconstruction locates the signal within the patient by using the Fourier transformation, which is the mathematical operation necessary to break the signal down into its constituents. To select a slice, two steps are necessary: (1) a slice-select gradient is imposed along an axis orthogonal to the plane of the desired slice and (2) an RF pulse with a limited range of frequencies or bandwidth is applied to excite only those nuclei that match the frequencies of precession [28].

The frequency and phase encoding are used to detect signals from each point (pixel) in the selected slice. In order to encode the imaged object along the so-called phase-encoding direction, a gradient along the direction of phase

encoding is applied, and thereafter the signal is sampled. To sample the whole object, it is necessary to apply this gradient several times, each time increasing the gradient by a fixed amount. The phase-encoding gradient (GPE) is turned on for a limited time period. While it is applied, it modifies the spin resonance frequency, inducing dephasing. Thereafter the GPE all the protons precess at the same frequency but with different phases. The protons in the same row, orthogonal to the GPE direction, have the same phase. The frequency encoding is the final step used to differentiate pixels with the same phase encoding, i.e., a single row of the object being imaged. A magnetic gradient during readout of the signal results in a specific shift of the precessing frequency in the horizontal direction throughout the time it is applied. Phase and frequency information are combined to fill in a matrix or grid. This grid is called the K-space. Finally, a fast Fourier transform transforms the raw data in a common image [26].

Key Learning Points

- Radio-frequency (RF) energy pulses of exactly the Larmor frequency are used to modify the net magnetization vector direction.
- After the Radio-frequency $\mathbf{B}_1$ is turned off, the net magnetization **vector** tends to realign itself with the external magnetic field $\mathbf{B}_0$. This event is referred to as free induction decay (FID).
- The time necessary to recover the thermal equilibrium is the **relaxation time**, which is constant for a given substance at a given magnetic field strength.
- Gradients, i.e, additional linear variations of the magnetic static field strength in a selected region, are employed to localize the MR signal in a region of interest.

8.4.4 Weightings and Image Contrast

Soft tissue contrast in MRI is principally due to the differences among the tissues' characteristic relaxation times. The type of pulse sequence and its parameters determines whether the image contrast is weighted by the T_1 or T_2 relaxation processes. To briefly summarize, two main controls determine tissue contrast in MRI: TR and TE. The echo time, TE, is the time between the application of the radio-frequency pulse and the beginning of NMR signal sampling. Meanwhile, the magnetization and the signal will diminish due to T_2-relaxation. The echo time controls the T_2-weighting in the images. A long TE compared to T_2 will thus result in considerable T_2-contrast, despite a little signal.

Image formation requires that similar measurements are repeated several times, e.g., once per line in the image. The repetition time, TR, is the length of time between two succeeding similar measurements or pulse sequence. TR controls the amount of longitudinal magnetization that recovers between each pulse. Indeed, each measurement makes use of the longitudinal magnetization present. Short repetition time will result in magnetization not having fully relaxed back into alignment before the next pulse is made, hence every repeat will therefore only produce a small signal. A long repetition time allows the magnetization to relax back into alignment with the main magnetic field. Short or long TR is relative to the maximum T_1 of tissues present in the sample. If TR is significantly longer than the maximum T_1, then magnetization reaches complete equilibrium, and the T_1 contrast disappears. Thus, using a long TR results in limited T_1-weighting but a strong signal. Conversely, the choice of applying short TR means that the signal is reduced for all types of tissue, but the signal becomes more T_1-weighted; in other words, the images will lose intensity but will gain greater signal variation between tissues with different T_1.

Finally, both T_1-contrast and T_2-contrast may be minimized, thus producing an image whose brightness is determined by the number of protons present in a voxel, also known as proton density (PD). In such an image, variation in the water content is the primary source of contrast. Short TR and TE are commonly chosen to acquire T_1-weighted images, since T_1 contrast is maximized and T_2 contrast is minimized. T_2-weighted images are produced if long TR and long TE are chosen, since T_1 contrast is minimized and T_2 contrast maximized. Finally, PD-weighted images are obtained by applying long TR and short TE, since both T_1- and T_2-contrast are minimized [23].

8.4.5 Sequences

An MRI sequence is an ordered combination of RF and gradient pulses designed to acquire the data to form the image. Several steps are required to acquire the data needed to create an MR image. First an RF pulse excites tissue magnetization in the presence of a slice-select gradient. Next, phase encoding and frequency encoding/readout spatially localize the protons in the other two dimensions. Finally, after the data have been collected, the process is repeated for a series of phase-encoding steps.

Spin echo (SE) and gradient-recalled echo (GRE) are the two fundamental types of MR pulse sequences. All other MR sequences are variations of these, with different parameters added on. MR pulse sequences can be either two-dimensional (2D), with one section acquired at a time, or three-dimensional (3D), with a volume of multiple sections obtained in a single acquisition.

In SE sequences, a 90° RF pulse flips the net magnetization **M** in the transverse plane. As the spinning nuclei experience relaxation, the transverse magnetization is gradually dephased. Thus, in order to rephase the spinning nuclei, a 180° RF pulse is applied at the time equal to one half the TE. Thus, at the total TE, the nuclei are spinning in phase and an echo is produced and read. In the GRE sequence, an RF pulse is applied to partly flip the net magnetization **M** into the transverse plane (variable flip angle). Gradients, instead of RF pulses, are exploited to dephase (negative gradient) and rephase (positive gradient) transverse magnetization [29].

Key Learning Points

- Soft tissue contrast in MRI is mainly due to the differences among the tissues' characteristic relaxation times.
- The echo time, TE, is the time between the application of the radio-frequency pulse and the beginning of NMR signal sampling; longer TE results in considerable T_2-contrast.
- The repetition time, TR, is the length of time between two succeeding similar measurements or pulse sequence; a long TR results in limited T_1-weighting but in a strong signal.
- An MRI sequence is an ordered combination of RF and gradient pulses designed to acquire the data to form the image.

8.4.6 MRI Hardware

8.4.6.1 Magnet

The hardware of the MRI scanner consists of four main components. The first component is the powerful magnet generating the strong magnetic field $\mathbf{B}_0$ essential to induce the net magnetization **M** in the body. Clinical studies require high image quality. This means that high field strength is preferred. Superconducting magnets capable of producing at least 1.5 Tesla are usually employed and, for some imaging studies, 3 Tesla instruments are preferred.

To eliminate the electrical resistance, the solenoid coils of niobium-titanium (NbTi) are enveloped by liquid helium (−269 °C). When NbTi becomes superconducting, the current will circulate indefinitely without any external power supply [30]. Iron-cored or permanent magnets have a more patient friendly design, whereas superconducting MRI system cryostats are usually shaped as a cylindrical tube that surrounds the patient. Although the cryostat bore is 100 cm wide, additional coils must be accomodated inside the bore; hence the diameter is reduced to 60 cm. Recently manufacturers have introduced system with 70 cm apertures and reduced magnet length to improve patient acceptance. In order to provide high-quality images, MRI magnets must produce a magnetic field with very high temporal and spatial uniformity, on the order of several parts-per-million (ppm) over the whole imaging volume. Typically, the 1.5 and 3 T magnets commercially available can guarantee field uniformity on the order of 10 ppm peak-to-peak in about 50 cm diameter volume. Magnetic shimming is necessary to improve the magnet uniformity to the design value of 10 ppm over the image volume. MRI systems are shimmed with two shimming methods: first, active shimming that makes use of superconducting coils placed in the cryostat. Second, passive shimming that adopts small pieces of iron located in the magnet bore. The magnet must be shimmed whenever it is moved to a new site because both shimming methods are site-dependent.

8.4.6.2 Gradients

The second component of MRI systems is a set of additional coils of wire located inside the magnetic bore, designed to create linear magnetic gradients in the three orthogonal physical directions x, y, and z. Gradient pulses are used to add a spatial variation to $\mathbf{B}_0$. Thus, spins at different physical locations will precess at different frequencies. As a result, the resonant frequencies of the hydrogen nuclei are spatially dependent within the gradient. At the middle of gradient, there is the null, where no change in precessional frequency occurs. A linear increase (or decrease) in precessional frequency exists with the variation of local magnetic field strength away from the null.

Location of spins along the gradient is determined by their frequency and phase. Often gradient pulses are trapezoidal in shape, since it takes time for the gradient field to ramp up to the desired amplitude and then to ramp down again. Gradient amplitudes are commonly measured in milliteslas per meter (mT/m); typical gradient amplitudes are in the range 30–50 mT/m. The time required for the waveform to ramp up and down is known as rise time, which is typically in the range of 200–1000 μs.

High-gradient amplitudes are useful to acquire thin slices or a small field of view (FOV), whereas short rise time permits to select short echo time, echo-spacing, and repetition time. Limitations on the maximum amplitude and rise time must be obeyed to avoid possible peripheral nerve stimulation, pain, and other effects. Gradient switching is responsible for the typical knocking noise heard during MRI examinations [31].

8.4.6.3 Radio-Frequency System

The third component of an MRI system is the apparatus that generates the radio frequency necessary to tip the net magnetization **M**.

The radio-frequency (RF) system is made up of a transmitter, coil, and receiver. The transmitter is in charge of generating suitably shaped pulses of current centered at the Larmor

frequency. When these pulses are applied to the coil, the alternating $\mathbf{B}_1$ field is produced. The coil will also detect the MR signal from the patient. The transmitter must generate RF pulses with appropriate center frequencies, bandwidths, amplitudes, and phases in order to excite spins within the desired slices or slabs. Setting the center frequency of the pulse is crucial to tell the scanner at exactly what frequency the nuclei of interest are resonating in the magnet's isocenter.

Setting the center frequency also allows identification and tuning on a nucleus of interest. The bandwidth, or the range of frequencies within the pulse, controls the thickness of the excited slice. The shape and duration of the RF pulse envelope determines the bandwidth. The amplitude of the RF pulse controls how much the magnetization is flipped by the pulse, whilst the phase controls along which axis the magnetization is flipped (in the rotating frame of reference). The main transmitting coil is usually the body coil, which surrounds the entire patient. This is usually built into the scanner bore and is not generally visible. Since this coil is large it has a very uniform transmission field, but this also means that it is not particularly sensitive if used as a receiver coil" [32].

A receiver coil (or simply "coil") has the function of maximizing signal detection, while minimizing the noise. The tissues of the patients are the principal source of noise. A method to reduce the noise and maximize the SNR is minimizing the coil dimensions, i.e., the coil's volume should be filled as much as possible by the sample; hence, the large number of anatomically optimized receiver coils available (e.g., head, spine, shoulder, breast, knee). The signal from each individual coil element is amplified and then digitized in preparation for reconstruction.

Two types of receiver coil are currently exploited: volume and surface. Volume coils completely surround the anatomy of interest and are often combined transmit/receive coils. Surface coils are generally receiver only and, as the name suggests, are good for detecting signal near the surface of the patient.

8.4.6.4 Computer Systems

Finally, the fourth component consists of a number of computers. The multitasking nature of MRI makes impractical to control the many processes from the host computer, so each subsystems will have dedicated microprocessors that download commands from the host.

Typically, MRI systems have a host computer on which the operator adjusts the scan parameters. These parameters are converted into commands and then are transferred to the pulse programmer (PP) (another microprocessor system) that controls the hardware. The PP manages the timing of RF, gradients, and data acquisition in such a way that they are all properly synchronized. As the data acquisition is completed, a separate computer system, known as the array processor, implements the image reconstruction. The reconstructed images are then passed back to the host for image display, processing, archiving, and networking.

Key Learning Points

- The main component of an MR system is the powerful magnet generating the strong magnetic field $\mathbf{B}_0$ necessary to induce the net magnetization $\mathbf{M}$ in the body.
- Gradients are a set of additional coils of wire located inside the magnetic bore, designed to create linear magnetic gradients in the three orthogonal directions *x, y, z*.
- The radio-frequency (RF) system is made up of a transmitter, a coil, and a receiver; the transmitter generates suitably shaped pulses of current centered at the Larmor frequency.
- MRI systems have a host computer employed to adjust the scan parameters.

8.4.7 MR Safety

This section briefly describes the potential bio-effects and risks of MRI environment for patients and personnel.

8.4.7.1 Projectile Risk

The static magnetic field $\mathbf{B}_0$ of an MRI scanner attracts ferromagnetic objects and accelerates them toward the center of the bore of the MRI scanner. Potential dangerous projectiles that can be torqued or, in the worst case, accelerated by $\mathbf{B}_0$ are objects such as coins, hairpins, steel oxygen tanks, or scissors. Every facility needs to warn personnel about the potential hazard and risks present in an MRI environment. Insufficient MRI safety training of ancillary medical personnel has led to fatal accidents [33].

All patients undergoing MRI exams must be screened for metal, medication patches, tattoos, body piercings, and any electrically, magnetically, or mechanically activated devices. Specifically trained personnel must perform the screening. The screening may be carried out by means of questionnaires, and it is fundamental to detect any implanted or external metallic foreign bodies. Magnetic field strength compatibility of implanted devices must be checked, possibly using device identification card, medical record, manufacturers' websites, and MRIsaftey.com [34].

8.4.7.2 Tissue Heating from RF

The main biological effects related to the exposure to RF fields are due to the heat generated in the body during the interaction with the electric field component of RF.

The specific absorption rate (SAR) is commonly used to describe the potential for heating of the patient's tis-

sues; this parameter is a measure of the power absorbed in tissue per unit mass. SAR is proportional to the product of the conductivity and the square of the electric field. The FDA and IEC limit SAR by anatomical site based on the potential effects of heating. In most countries standard MRI systems are limited to a maximum SAR of 4 W/kg [35].

Key Learning Points

- The static magnetic field $\mathbf{B}_0$ of an MRI scanner attracts ferromagnetic objects and accelerates them toward the center of the bore of the scanner.
- Patients undergoing MRI scans must be screened for metal, medication patches, tattoos, body piercings, and any electrically, magnetically, or mechanically activated devices.
- The main biological effects related to the exposure to RF fields are due to the heat generated in the body during the interaction with the electric field component of RFs.

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