Magnetic Resonance Imaging (MRI)

MRI

MRI is a noninvasive cross-sectional imaging modality that does not require any ionizing radiation. For acquiring images, MRI uses the physical principle of magnetic resonance that was first described by Felix Bloch and Edward Purcell in 1946 who then received the Nobel Prize in Physics in 1952 for their discovery.

MR Imaging Principle

A patient will be moved inside the bore of a magnet with a strong static magnetic field (B_o) that is in the range of 0.2T- 3T for clinical MRI scanners (Fig.1). For the typical clinical magnetic field strengths of 1.5T and 3T, strong superconducting magnets are required.

The human body is composed of water molecules, which contain two hydrogen nuclei, or protons. The magnetic moments of these protons align with the direction of the static magnetic field inside the scanner. Oscillatin; electromagnetic radiofrequency fields and gradient fields are then used to acquire images from the body of the patient. The radiofrequency field (B_1) is produced by an (RF) coil, and the fast-switching gradient fields are produced by three different coil systems (G_x, G_y, and G_z) that are embedded in the bore of the MRI scanner.

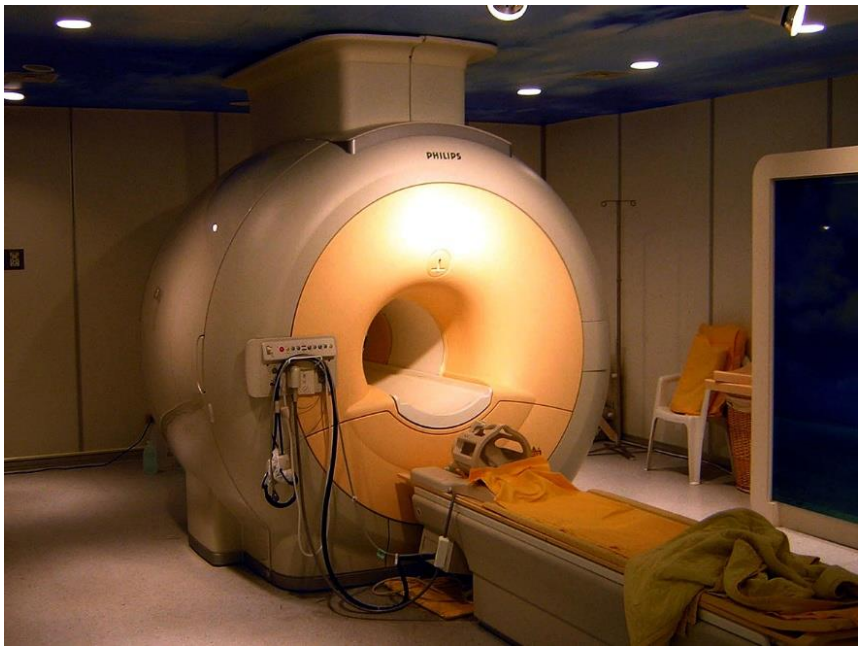

Fig. 1 Clinical MRI system with the patient table in the front that can be moved inside the bore of the magnet.

Static Magnetic Field (B_o)

The strong static magnetic field (B_o) is used to prepare chemical compounds (mainly hydrogen protons in water and fat) for imaging. The strength of the static magnetic field B0 of an MR scanner, the magnetic flux density, is measured in the SI unit Tesla (T). Stronger static magnetic fields lead to a higher signal-to-noise ratio (SNR) and subsequently to a better image quality in the MR images or to a faster scan time. The static magnetic field of a 3 T MRI scanner is approximately 60,000 times stronger than the magnetic field of the Earth (∽50 μT).

These high magnetic fields cannot be achieved with permanent magnets and require superconducting magnets. A superconducting magnet is an electromagnet that is made from superconducting wires that are cooled with liquid helium (Fig.2). For clinical MRI systems, these superconducting wires are most commonly made of an alloy of niobium and titanium (NbTi). The wires will be cooled below their critical temperature, the temperature at which the winding material changes from the normal resistive state to a superconducting state. The wires can conduct large electric currents in the superconducting state and have zero electrical resistance to produce strong magnet fields.

Fig. 2 Superconducting magnet of a clinical MRI system after the removal of the covers and the patient table.

Radiofrequency Field (B_1)

Radiofrequency (RF) coils are designed as antennas that can transmit radiofrequency waves inside the human body and also receive the radiofrequency waves from the human body. The radiofrequency coils are built for different body parts of the patient (e.g., head coils, abdominal coils, knee coils, extremity coils) and are positioned as close as possible to the anatomical structures of interest to achieve a good image quality. A dedicated head coil is shown in Fig. 1 at the head end of the patient table.

The radiofrequency field B_1 causes the protons to alter their alignment relative to the static magnetic field B_0 at a higher energy state. This process is called excitation.

The protons that were excited by the B_1 field will then return to their lower energy state in a process called relaxation and will reemit RF radiation at the Larmor frequency:

$$\omega_0 = \gamma B_0$$

Returning RF waves from the patient are picked up by the RF coils, stored in an intermediate image space (k-space), and then used to calculate an MR image with the mathematical fast Fourier transform (FFT).

Gradient Magnetic Fields (G_x, G_y, and G_z)

Gradient fields are used to localize the MR signal from a specific location in the human body. The three different gradient coils G_x, G_y, and G_z are embedded inside the bore of the MRI scanner and are used for slice selection (z), phase encoding (x), and frequency encoding (y) to acquire cross-sectional 2D images from any angulation or 3D datasets of the human body (Fig. 3). The gradient coils are responsible for the acoustic noise in an MRI scanner caused by magnetic Lorentz forces from the static magnetic field (B_o) on the electric currents flowing in the gradient coils.

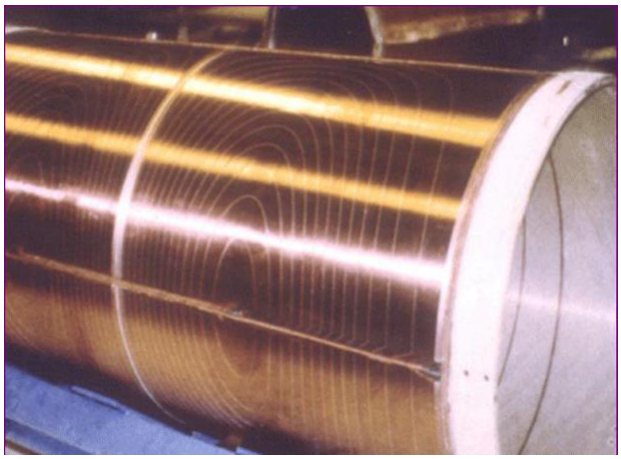

Fig. 3 Gradient system of an MRI system with x-, y-, and z-gradient coils embedded in epoxy. The gradient coils are used to localize the MR signal from a specific location in the human body.

MRI Pulse Sequences

MRI pulse sequences are programs that contain the timing and duration of radiofrequency pulses and magnetic gradients to produce an image. In MRI there are two major pulse sequence groups: spin echo (SE) sequences and gradient echo (GRE) sequences. Spin echo sequences include a slice selective 90° excitation pulse followed by one or more 180° refocusing pulses (Fig. 4). Gradient echo sequences are characterized by the use of excitation pulses with flip angles of usually less than 90°, the absence of 180° RF refocusing pulses, and the use of dephasing and rephrasing gradient pulses (Fig. 5). Gradient echo sequences are in general faster than spin echo sequences and allow real-time imaging of moving organs in the human body such as the heart.

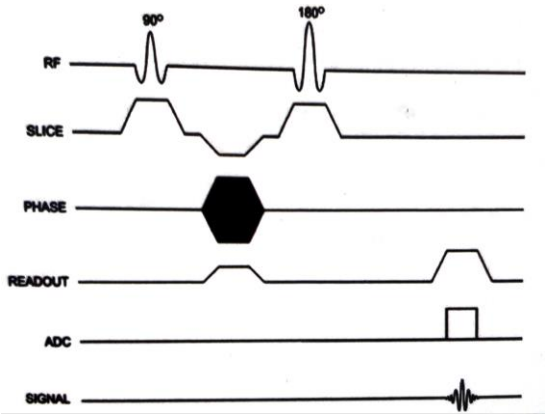

Fig. 4 Spin echo (SE) pulse sequence diagram with a slice selective 90 excitation pulse followed by a 180° refocusing pulses and the gradients in slice, phase, and frequency/readout direction. ADC is the time of signal acquisition and analog-to-digital signal conversion

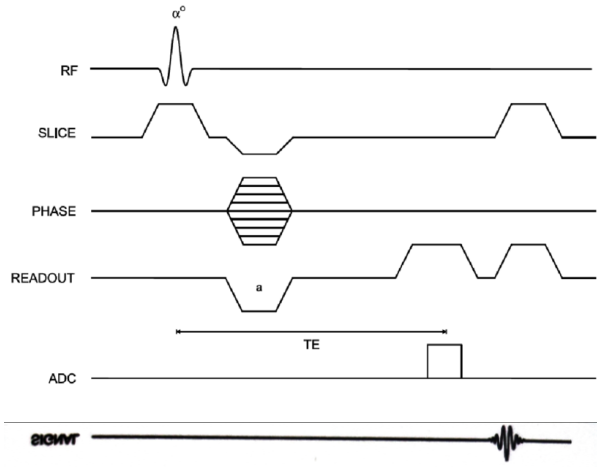

Fig. 5 Gradient echo (GRE) pulse sequence diagram with a slice selective excitation pulse a, which is typically smaller than 90. The gradients in readout direction are used to produce a gradient echo when the ADC is turned on during signal acquisition.

$$S = \rho . \left(1 - e^{-\frac{TR}{T_1}}\right) e^{-\frac{TR}{T_2}}$$

The repetition time TR is the time between two excitation pulses. The echo time (TE) is the time between an excitation pulse and MR signal sampling when the echo maximum occurs (Fig. 4). In a gradient echo sequence, there are in general three parameters, that influence the contrast: the flip angle of the excitation pulse t, TE and TR(Fig.5)