

NUCLEAR MAGNETIC RESONANCE

Nuclear magnetic resonance (NMR) is the spectroscopic study of the magnetic properties of the *nucleus* of the atom. The protons and neutrons of the nucleus have a *magnetic* field associated with their nuclear spin and charge distribution. *Resonance* is an energy coupling that causes the individual nuclei, when placed in a strong external magnetic field, to selectively absorb, and later release, energy unique to those nuclei and their surrounding environment. The detection and analysis of the NMR signal has been extensively studied since the 1940s as an analytic tool in chemistry and biochemistry research. NMR is not an imaging technique but rather a method to provide spectroscopic data concerning a sample placed in the device. In the early 1970s, it was realized that magnetic field gradients could be used to localize the NMR signal and to generate images that display magnetic properties of the proton, reflecting clinically relevant information. As clinical imaging applications increased in the mid-1980s, the “nuclear” connotation was dropped, and magnetic resonance imaging (MRI), with a plethora of associated acronyms, became commonly accepted in the medical community.

MRI is a rapidly changing and growing image modality. The high contrast sensitivity to soft tissue differences and the inherent safety to the patient resulting from the use of nonionizing radiation have been key reasons why MRI has supplanted many CT and projection radiography methods. With continuous improvements in image quality, acquisition methods, and equipment design, MRI is the modality of choice to examine anatomic and physiologic properties of the patient. There are drawbacks, including high equipment and siting costs, scan acquisition complexity, relatively long imaging times, significant image artifacts, and patient claustrophobia problems.

The expanding use of MRI demands an understanding of the underlying MR principles to glean the most out of the modality. This chapter reviews the basic properties of magnetism, concepts of resonance, and generation of tissue contrast through specific pulse sequences. Details of image formation, characteristics, and artifacts, as well as equipment operation and biologic effects, are discussed in Chapter 15.

14.1 MAGNETIZATION PROPERTIES

Magnetism

Magnetism is a fundamental property of matter; it is generated by moving charges, usually electrons. Magnetic properties of materials result from the organization and

motion of the electrons in either a random or a nonrandom alignment of magnetic “domains,” which are the smallest entities of magnetism. Atoms and molecules have electron orbitals that can be paired (an even number of electrons cancels the magnetic field) or unpaired (the magnetic field is present). Most materials do not exhibit overt magnetic properties, but one notable exception is the permanent magnet, in which the individual domains are aligned in one direction.

Magnetic susceptibility describes the extent to which a material becomes magnetized when placed in a magnetic field. The induced internal magnetization can oppose the external magnetic field and lower the local magnetic field surrounding the material. On the other hand, the internal magnetization can form in the same direction as the applied magnetic field and increase the local magnetic field. Three categories of susceptibility are defined: *diamagnetic*, *paramagnetic*, and *ferromagnetic*. Diamagnetic materials have slightly negative susceptibility and oppose the applied magnetic field. Examples of diamagnetic materials are calcium, water, and most organic materials (chiefly owing to the diamagnetic characteristics of carbon and hydrogen). Paramagnetic materials have slightly positive susceptibility and enhance the local magnetic field, but they have no measurable self-magnetism. Examples of paramagnetic materials are molecular oxygen (O_2), some blood degradation products, and gadolinium-based contrast agents. Ferromagnetic materials are “superparamagnetic”—that is, they augment the external magnetic field substantially. These materials can exhibit “self-magnetism” in many cases. Examples are iron, cobalt, and nickel.

Unlike the monopole electric charges from which they are derived, magnetic fields exist as dipoles, where the north pole is the origin of the magnetic field lines and the south pole is the return (Fig. 14-1A). One pole cannot exist without the other. As with electric charges, “like” magnetic poles repel and “opposite” poles attract. The *magnetic field strength*, B , (also called the magnetic flux density) can be

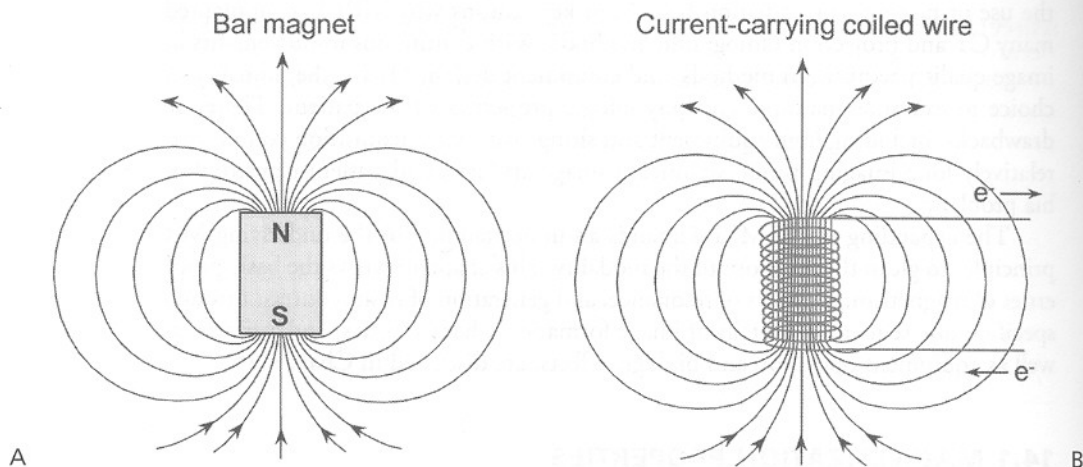


FIGURE 14-1. A: The magnetic field has two poles, with magnetic field lines emerging from the north pole (N) and returning to the south pole (S), as illustrated by a simple bar magnet. **B:** A coiled wire carrying an electric current produces a magnetic field with characteristics similar to those of a bar magnet. Magnetic field strength and field density depend on the magnitude of the current and the number of coil turns. Note: the flow of positive current is opposite that of the electron flow.

conceptualized as the number of magnetic lines of force per unit area. The magnetic field drops off with the square of the distance. The SI unit for B is the tesla (T), and as a benchmark, the earth's magnetic field is about $1/20,000$ T. An alternate unit is the gauss (G), where $1 \text{ T} = 10,000 \text{ G}$.

Magnetic fields can be induced by a moving charge in a wire (for example, see the section on transformers in Chapter 5). The direction of the magnetic field depends on the sign and the direction of the charge in the wire, as described by the "right hand rule": The fingers point in the direction of the magnetic field when the thumb points in the direction of a moving positive charge (i.e., opposite the direction of electron movement). Wrapping the current-carrying wire many times in a coil causes a superimposition of the magnetic fields, augmenting the overall strength of the magnetic field inside the coil, with a rapid falloff of field strength outside the coil (see Fig. 14-1B). Direct current (DC) amplitude in the coil determines the overall magnitude of the magnetic field strength. In essence, this is the basic design of the "air core" magnets used for diagnostic imaging, which have magnetic field strengths ranging from 0.3 to 2.0 T, where the strength is directly related to the current.

Magnetic Characteristics of the Nucleus

The nucleus exhibits magnetic characteristics on a much smaller scale as described in the previous section for atoms and molecules and their associated electron distributions. The nucleus is comprised of protons and neutrons with characteristics listed in Table 14-1. Magnetic properties are influenced by the spin and charge distributions intrinsic to the proton and neutron. For the proton, which has a unit positive charge (equal to the electron charge but of opposite sign), the nuclear "spin" produces a magnetic dipole. Even though the neutron is electrically uncharged, charge inhomogeneities on the subnuclear scale result in a magnetic field of opposite direction and of approximately the same strength as the proton. The *magnetic moment*, represented as a vector indicating magnitude and direction, describes the magnetic field characteristics of the nucleus. A phenomenon known as *pairing* occurs within the nucleus of the atom, where the constituent protons and neutrons determine the nuclear magnetic moment. If the total number of protons (P) and neutrons (N) in the nucleus is even, the magnetic moment is essentially zero. However, if N is even and P is odd, or N is odd and P is even, the noninteger nuclear spin generates a magnetic moment. A single atom does not generate a large enough nuclear magnetic moment to be observable; the signal measured by an MRI system is the conglomerate signal of billions of atoms.

TABLE 14-1. PROPERTIES OF THE NEUTRON AND PROTON

Characteristic	Neutron	Proton
Mass (kg)	1.674×10^{-27}	1.672×10^{-27}
Charge (coulomb)	0	1.602×10^{-19}
Spin quantum number	$\frac{1}{2}$	$\frac{1}{2}$
Magnetic moment (joule/tesla)	-9.66×10^{-27}	1.41×10^{-26}
Magnetic moment (nuclear magneton)	-1.91	2.79

Nuclear Magnetic Characteristics of the Elements

Biologically relevant elements that are candidates for producing MR images are listed in Table 14-2. The key features include the strength of the magnetic moment, the physiologic concentration, and the isotopic abundance. Hydrogen, having the largest magnetic moment and greatest abundance, is by far the best element for general clinical utility. Other elements are orders of magnitude less sensitive when the magnetic moment and the physiologic concentration are considered together. Of these, ^{23}Na and ^{31}P have been used for imaging in limited situations, despite their relatively low sensitivity. Therefore, the proton is the principal element used for MR imaging.

The spinning proton or “spin” (spin and proton are synonymous herein) is classically considered to be like a bar magnet with north and south poles (Fig. 14-2); however, the magnetic moment of a single proton is extremely small and not detectable. A vector representation (amplitude and direction) is helpful when contemplating the additive effects of many protons. Thermal energy agitates and randomizes the direction of the spins in the tissue sample, and as a result there is no net tissue magnetization (Fig. 14-3A). Under the influence of a strong external magnetic field, B_0 , however, the spins are distributed into two energy states: alignment with (parallel to) the applied field at a low-energy level, and alignment against (antiparallel to) the field at a slightly higher energy level (see Fig. 14-3B). A slight majority of spins exist in the low-energy state, the number of which is determined by the thermal energy of the sample (at absolute zero, 0 degrees Kelvin (K), all protons would be aligned in the low-energy state). For higher applied magnetic field strength, the energy separation of the low and high energy levels is greater, as is the number of excess protons in the low-energy state. The number of excess protons in the low-energy state at 1.0 T is about 3 spins per million (3×10^{-6}) at physiologic temperatures and is proportional to the external magnetic field. Although this does not seem significant, for a typical voxel volume in MRI there are about 10^{21} protons, so there are $3 \times 10^{-6} \times 10^{21}$, or approximately 3×10^{15} , more spins in the low-energy state! This number of protons produces an observable magnetic moment when summed.

In addition to energy separation of the spin states, the protons also experience a torque from the applied magnetic field that causes precession, in much the same

TABLE 14-2. MAGNETIC RESONANCE PROPERTIES OF MEDICALLY USEFUL NUCLEI

Nucleus	Spin Quantum Number	% Isotopic Abundance	Magnetic Moment	Relative Physiologic Concentration*	Relative Sensitivity
^1H	$\frac{1}{2}$	99.98	2.79	100	1
^{16}O	0	99.0	0	50	0
^{17}O	$\frac{5}{2}$	0.04	1.89	50	9×10^{-6}
^{19}F	$\frac{1}{2}$	100	2.63	4×10^{-6}	3×10^{-8}
^{23}Na	$\frac{3}{2}$	100	2.22	8×10^{-2}	1×10^{-4}
^{31}P	$\frac{1}{2}$	100	1.13	7.5×10^{-2}	6×10^{-5}

*Note: all isotopes of a given element.

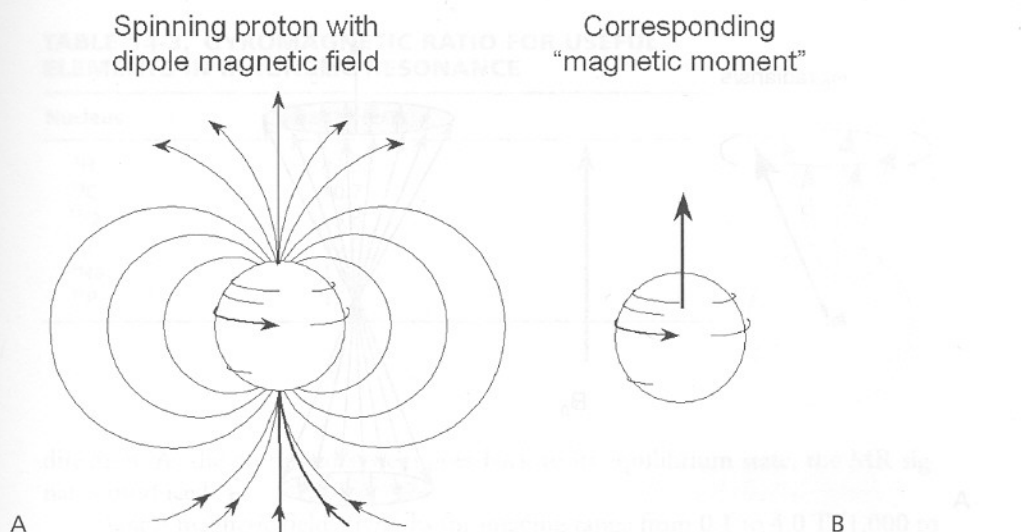


FIGURE 14-2. A: A spinning proton can classically be represented with a dipole magnetic field. **B:** Magnetic characteristics of the proton are described as a magnetic moment, with a vector that indicates direction and magnitude.

way that a spinning top wobbles due to the force of gravity (Fig. 14-4A). Direction of the spin axis is perpendicular to the torque's twisting. This precession occurs at an angular frequency (number of rotations/sec about an axis of rotation) that is proportional to the magnetic field strength B_0 . The *Larmor equation* describes the dependence between the magnetic field, B_0 , and the precessional angular frequency, ω_0 :

$$\omega_0 = \gamma B_0$$

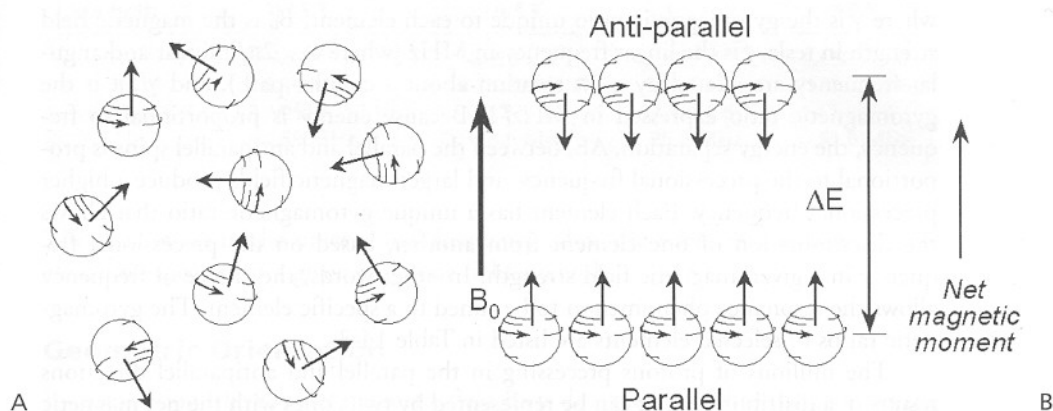


FIGURE 14-3. Simplified distributions of "free" protons without and with an external magnetic field are shown. **A:** Without an external magnetic field, a group of protons assumes a random orientation of magnetic moments, producing an overall magnetic moment of zero. **B:** Under the influence of an applied external magnetic field, B_0 , the protons assume a nonrandom alignment in two possible orientations: parallel and antiparallel to the applied magnetic field. A slightly greater number of protons exist in the parallel direction, resulting in a measurable *sample magnetic moment* in the direction of B_0 .

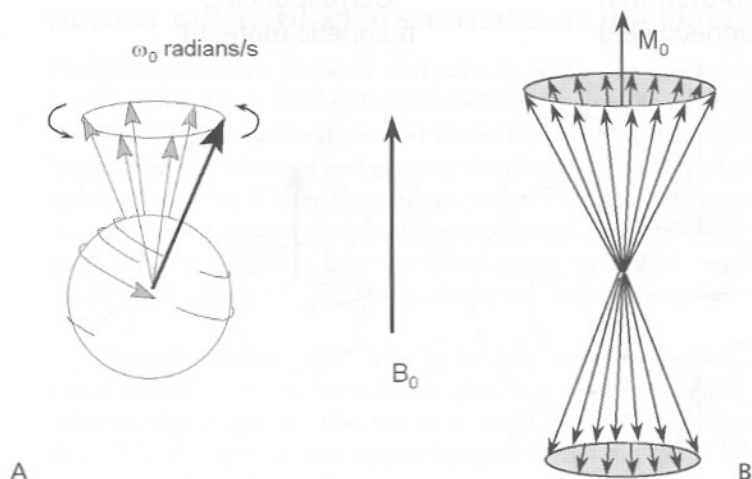


FIGURE 14-4. A: A single proton *precesses* about its axis with an angular frequency, ω , that is proportional to the externally applied magnetic field strength, according to the Larmor equation. **B:** A group of protons in the parallel and antiparallel energy states generates an equilibrium magnetization, M_0 , in the direction of the applied magnetic field B_0 . The protons are distributed randomly over the surface of the cone and produce no magnetization in the perpendicular direction.

or, with respect to linear frequency:

$$f_0 = \frac{\gamma}{2\pi} B_0$$

where γ is the gyromagnetic ratio unique to each element, B_0 is the magnetic field strength in tesla, f is the linear frequency in MHz (where $\omega = 2\pi f$: linear and angular frequency are related by a 2π rotation about a circular path), and $\gamma/2\pi$ is the gyromagnetic ratio expressed in MHz/T. Because energy is proportional to frequency, the energy separation, ΔE , between the parallel and antiparallel spins is proportional to the precessional frequency, and larger magnetic fields produce a higher precessional frequency. Each element has a unique gyromagnetic ratio that allows the discrimination of one element from another, based on the precessional frequency in a given magnetic field strength. In other words, the choice of frequency allows the resonance phenomenon to be tuned to a specific element. The gyromagnetic ratios of selected elements are listed in Table 14-3.

The millions of protons precessing in the parallel and antiparallel directions results in a distribution that can be represented by two cones with the net magnetic moment equal to the vector sum of all the protons in the sample in the direction of the applied magnetic field (see Fig. 14-4B). At equilibrium, no magnetic field exists perpendicular to the direction of the external magnetic field because the individual protons precess with a random distribution, which effectively averages out any net magnetic moment. Energy (in the form of a pulse of radiofrequency electromagnetic radiation) at the precessional frequency (related to ΔE) is absorbed and converts spins from the low-energy, parallel direction to the higher-energy, antiparallel

TABLE 14-3. GYROMAGNETIC RATIO FOR USEFUL ELEMENTS IN MAGNETIC RESONANCE

Nucleus	$\gamma/2\pi$ (MHz/T)
^1H	42.58
^{13}C	10.7
^{17}O	5.8
^{19}F	40.0
^{23}Na	11.3
^{31}P	17.2

direction. As the perturbed system goes back to its equilibrium state, the MR signal is produced.

Typical magnetic field strengths for imaging range from 0.1 to 4.0 T (1,000 to 40,000 G). For protons, the precessional frequency is 42.58 MHz in a 1-T magnetic field (i.e., $\gamma/2\pi = 42.58$ MHz/T for ^1H). The frequency increases or decreases linearly with increases or decreases in magnetic field strength, as shown in the example. Accuracy and precision are crucial for the selective excitation of a given nucleus in a magnetic field of known strength. Spin precession frequency must be known to an extremely small fraction (10^{-12}) of the precessional frequency for modern imaging systems.

Example: What is the frequency of precession of ^1H and ^{31}P at 0.15 T? 0.5 T? 1.5 T? 3.0 T?

The Larmor frequency is calculated as $f_0 = (\gamma/2\pi)B_0$.

Field Strength	0.15 T	0.5 T	1.5 T	3.0 T
^1H	$f = 42.58 \text{ MHz/T} \times 0.15 \text{ T}$ = 6.39 MHz	42.58×0.5 = 21.29 MHz	42.58×1.5 = 63.87 MHz	42.58×3.0 = 127.74 MHz
^{31}P	$f = 17.2 \text{ MHz/T} \times 0.15 \text{ T}$ = 2.58 MHz	17.2×0.5 = 8.6 MHz	17.2×1.5 = 25.8 MHz	17.2×3 = 51.6 MHz

The differences in the precessional frequency allow the selective excitation of one elemental species for a given magnetic field strength.

Geometric Orientation

By convention, the applied magnetic field B_0 is directed parallel to the z-axis of the three-dimensional Cartesian coordinate axis system. The x and y axes are perpendicular to the z direction. For convenience, two frames of reference are used: the *laboratory frame* and the *rotating frame*. The laboratory frame (Fig. 14-5A) is a stationary reference frame from the observer's point of view. The proton's magnetic moment precesses about the z-axis in a circular geometry about the x-y plane. The rotating frame (see Fig. 14-5B) is a *spinning* axis system whereby the angular frequency is equal to the precessional frequency of the protons. In this frame, the spins

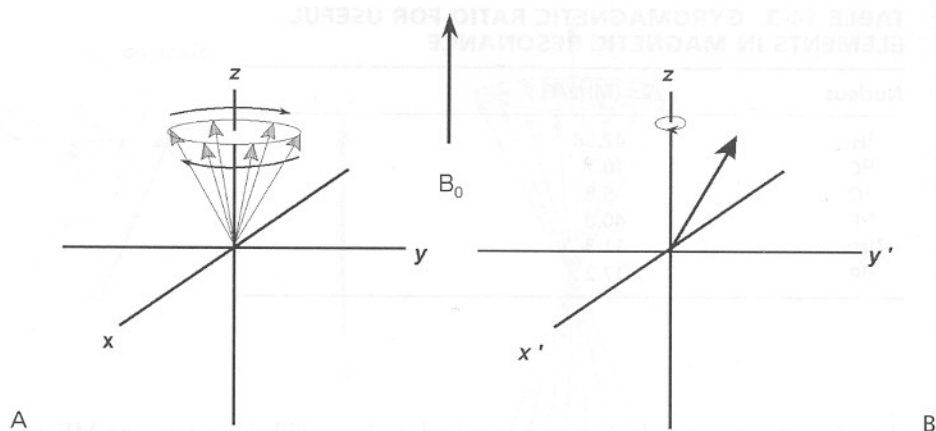


FIGURE 14-5. A: The laboratory frame of reference uses stationary three-dimensional Cartesian coordinates. The magnetic moment precesses around the z -axis at the Larmor frequency. **B:** The rotating frame of reference uses Cartesian coordinate axes that rotate about the z -axis at the Larmor precessional frequency, and the other axes are denoted x' and y' . When precessing at the Larmor frequency, the sample magnetic moment appears stationary.

appear to be stationary when they rotate at the precessional frequency. If a slightly higher precessional frequency occurs, a slow clockwise rotation is observed. For a slightly lower precessional frequency, counterclockwise rotation is observed. A merry-go-round exemplifies an analogy to the laboratory and rotating frames of reference. Externally, from the laboratory frame of reference, the merry-go-round rotates at a specific angular frequency (e.g., 15 rotations per minute [rpm]). Individuals riding the horses are observed moving in a circular path around the axis of rotation, and up-and-down on the horses. If the observer jumps *onto* the merry-go-round, everyone on the ride now appears stationary (with the exception of the up-and-down motion of the horses)—this is the *rotating* frame of reference. Even though the horses are moving up and down, the ability to study them in the rotating frame is significantly improved compared with the laboratory frame. If the merry-go-round consists of three concentric rings that rotate at 14, 15, and 16 rpm and the observer is on the 15-rpm section, all individuals on that particular ring would appear stationary, but individuals on the 14-rpm ring would appear to be rotating in one direction at a rate of 1 rpm, and individuals on the 16-rpm ring would appear to be rotating in the other direction at a rate of 1 rpm. Both the laboratory and the rotating frame of reference are useful in explaining various interactions of the protons with externally applied static and rotating magnetic fields.

The net magnetization vector, M , is described by three components. M_z is the component of the magnetic moment parallel to the applied magnetic field and is known as *longitudinal magnetization*. At equilibrium, the longitudinal magnetization is maximal and is denoted as M_0 , the equilibrium magnetization, where $M_0 = M_z$, with the amplitude determined by the excess number of protons that are in the low-energy state (i.e., aligned with B_0). M_{xy} is the component of the magnetic moment perpendicular to the applied magnetic field and is known as *transverse magnetization*. At equilibrium, the transverse magnetization is zero, because the vector components of the spins are randomly oriented about 360 degrees in the x - y

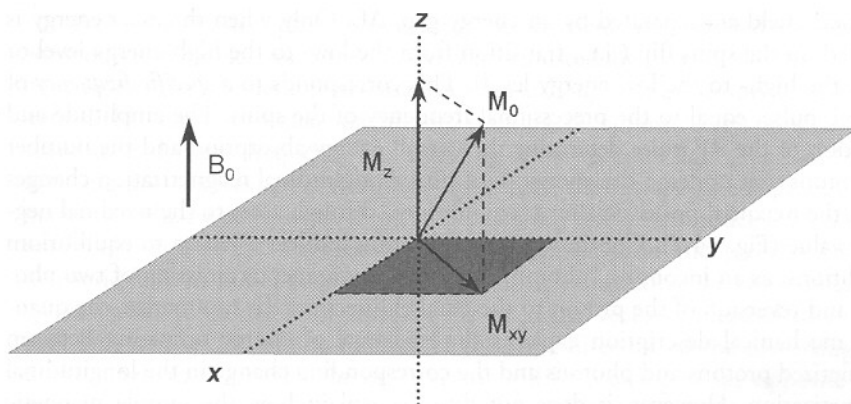


FIGURE 14-6. Longitudinal magnetization, M_z , is the vector component of the magnetic moment in the z direction. Transverse magnetization, M_{xy} , is the vector component of the magnetic moment in the x - y plane. Equilibrium magnetization, M_0 , is the maximal longitudinal magnetization of the sample; in this illustration it is shown displaced from the z -axis.

plane and cancel each other. When the system absorbs energy, M_z is “tipped” into the transverse plane; why this is important is explained in the next section. Figure 14-6 illustrates these concepts.

The famous *Bloch equations* are a set of coupled differential equations that describe the behavior of the magnetization vectors under any conditions. These equations, when integrated, yield the x' , y' , and z components of magnetization as a function of time.

14.2 GENERATION AND DETECTION OF THE MAGNETIC RESONANCE SIGNAL

Application of radiofrequency (RF) energy synchronized to the precessional frequency of the protons causes displacement of the tissue magnetic moment from equilibrium conditions (i.e., more protons are in the antiparallel orientation). Return to equilibrium results in emission of MR signals proportional to the number of excited protons in the sample, with a rate that depends on the characteristics of the tissues. Excitation, detection, and acquisition of the signals constitute the basic information necessary for MR spectroscopy and imaging.

Resonance and Excitation

The displacement of the equilibrium magnetization occurs when the magnetic component of the RF pulse, also known as the B_1 field, is precisely matched to the precessional frequency of the protons to produce a condition of *resonance*. This RF frequency (proportional to energy) corresponds to the energy separation between the protons in the parallel and antiparallel directions, as described by either a quantum mechanics or a classical physics approach. Each has its advantages in the understanding of MR physics.

The quantum mechanics approach considers the RF energy as photons (quanta) instead of waves. Spins oriented parallel and antiparallel to the external

magnetic field are separated by an energy gap, ΔE . Only when the *exact* energy is applied do the spins flip (i.e., transition from the low- to the high-energy level or from the high- to the low-energy level). This corresponds to a *specific frequency* of the RF pulse, equal to the precessional frequency of the spins. The amplitude and duration of the RF pulse determine the overall energy absorption and the number of protons that undergo the energy transition. Longitudinal magnetization changes from the maximal positive value at equilibrium, through zero, to the maximal negative value (Fig. 14-7). Continued RF application induces a return to equilibrium conditions, as an incoming "photon" causes the spontaneous emission of two photons and reversion of the proton to the parallel direction. To summarize, the quantum mechanical description explains the exchange of energy occurring between magnetized protons and photons and the corresponding change in the longitudinal magnetization. However, it does not directly explain how the sample magnetic moment induces a current in a coil and produces the MR signal. The classical physics model better explains this phenomenon.

Example: Determine the energy difference, ΔE (in eV), of the parallel and antiparallel spin states under the influence of a 1.5-T magnetic field and compare that value with an x-ray photon of 50 keV. Information needed includes the relationship between energy and wavelength— E (in keV) = $1.24/\lambda$ (in nm); between magnetic field strength and frequency— $\omega_0 = \gamma B_0$; and between frequency and wavelength— $\lambda = c/f$.

Answer: The precessional frequency, $\omega_0 = \gamma B_0 = 42.58 \text{ MHz/T} \times 1.5 \text{ T} = 63.86 \text{ MHz}$; $\lambda = c/f = (3.0 \times 10^8 \text{ m/sec}) / (63.86 \times 10^6 \text{ /sec}) = 4.7 \text{ m}$; $E = 1.24/\lambda = 1.24 / (4.69 \times 10^9) = 2.64 \times 10^{-10} \text{ keV} = 2.64 \times 10^{-7} \text{ eV} = \Delta E$. The 50-keV photon therefore possesses 1.9×10^8 more energy than the photon that causes a magnetized proton to flip in a 1.5-T magnetic field.

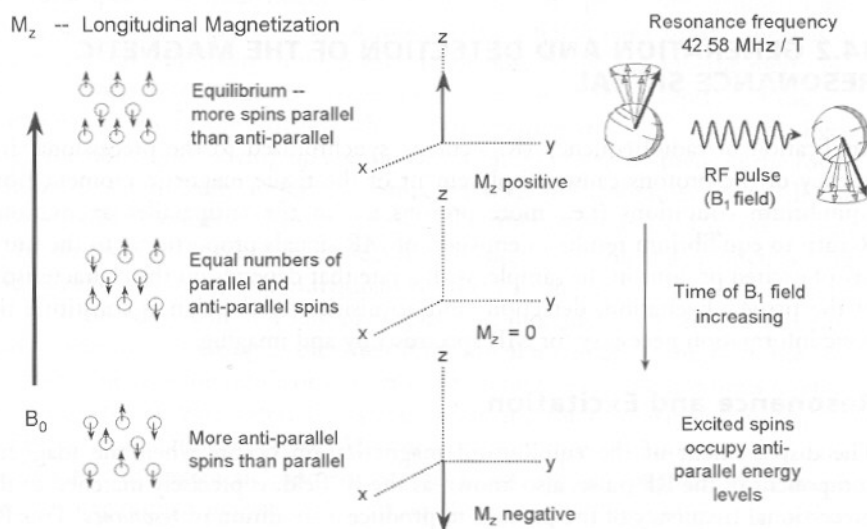


FIGURE 14-7. A simple quantum mechanics process depicts the discrete energy absorption and the time change of the longitudinal magnetization vector as radiofrequency (RF) energy, equal to the energy difference between the parallel and antiparallel spins, is applied to the sample (at the Larmor frequency). Discrete quanta absorption changes the proton energy from parallel to antiparallel. With continued application of RF energy at the Larmor frequency, M_z is displaced from equilibrium, through zero, to the opposite direction (high-energy state).

From the classical physics viewpoint, the B_1 field is considered the magnetic component of an electromagnetic wave with sinusoidally varying electric and magnetic fields. The magnetic field variation can be thought of as comprising two magnetic field vectors of equal magnitude, rotating in opposite directions around a point at the Larmor frequency and traveling at the speed of light (Fig. 14-8). At 0, 180, and 360 degrees about the circle of rotation, the vectors cancel, producing no magnetic field. At 90 degrees, the vectors positively add and produce the peak magnetic field, and at 270 degrees the vectors negatively add and produce the peak magnetic field in the opposite direction, thus demonstrating the characteristics of the magnetic variation. Now consider the magnetic vectors independently; of the two, only one vector will be rotating in the same direction as the precessing spins in the magnetized sample (the other vector will be rotating in the opposite direction). From the perspective of the *rotating frame*, this magnetic vector (the applied B_1 field) is stationary relative to the precessing protons within the sample, and it is applied along the x' -axis (or the y' -axis) in a direction *perpendicular* to the sample magnetic moment, M_z (Fig. 14-9A). The stationary B_1 field applies a torque to M_z , causing a rotation away from the longitudinal direction into the transverse plane. The rotation of M_z occurs at an angular frequency equal to $\omega_1 = \gamma B_1$. Because the tip angle, θ , is equal to $\omega_1 \times t$, where t is the time of the B_1 field application, then, by substitution, $\theta = \gamma B_1 t$, which shows that the time of the applied B_1 field determines the amount of rotation of M_z . Applying the B_1 field in the opposite direction (180-degree change in phase) changes the tip direction of the sample moment. If the RF energy is not applied at the precessional (Larmor) frequency, the B_1 field is not stationary in the rotating frame and the coupling (resonance) between the two magnetic fields does not occur (see Fig. 14-9B).

Flip angles describe the rotation through which the longitudinal magnetization vector is displaced to generate the transverse magnetization (Fig. 14-10). Common angles are 90 degrees ($\pi/2$) and 180 degrees (π), although a variety of smaller (less than 90 degrees) and larger angles are chosen to enhance tissue contrast in various

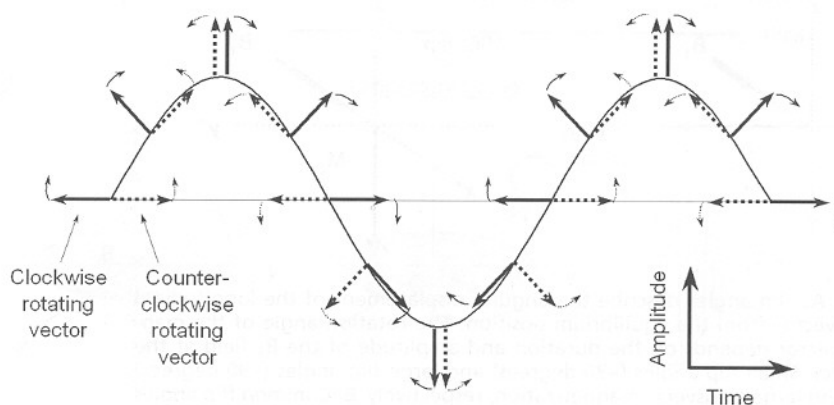


FIGURE 14-8. A classical physics description of the magnetic field component of the radiofrequency pulse (the electric field is not shown). Clockwise (*solid*) and counter-clockwise (*dotted*) rotating magnetic vectors produce the magnetic field variation by constructive and destructive interaction. At the Larmor frequency, one of the magnetic field vectors rotates synchronously in the rotating frame and is therefore stationary (the other vector rotates in the opposite direction).

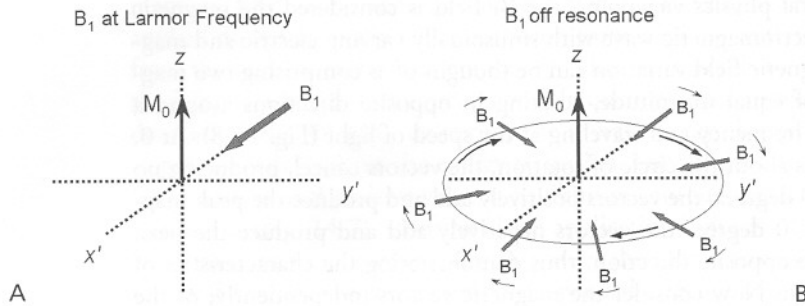


FIGURE 14-9. A: In the rotating frame, the radiofrequency pulse (B_1 field) is applied at the Larmor frequency and is stationary in the x' - y' plane. The B_1 field interacts at 90 degrees to the sample magnetic moment and produces a torque that displaces the magnetic vector away from equilibrium. **B:** The B_1 field is not tuned to the Larmor frequency and is not stationary in the rotating frame. No interaction with the sample magnetic moment occurs.

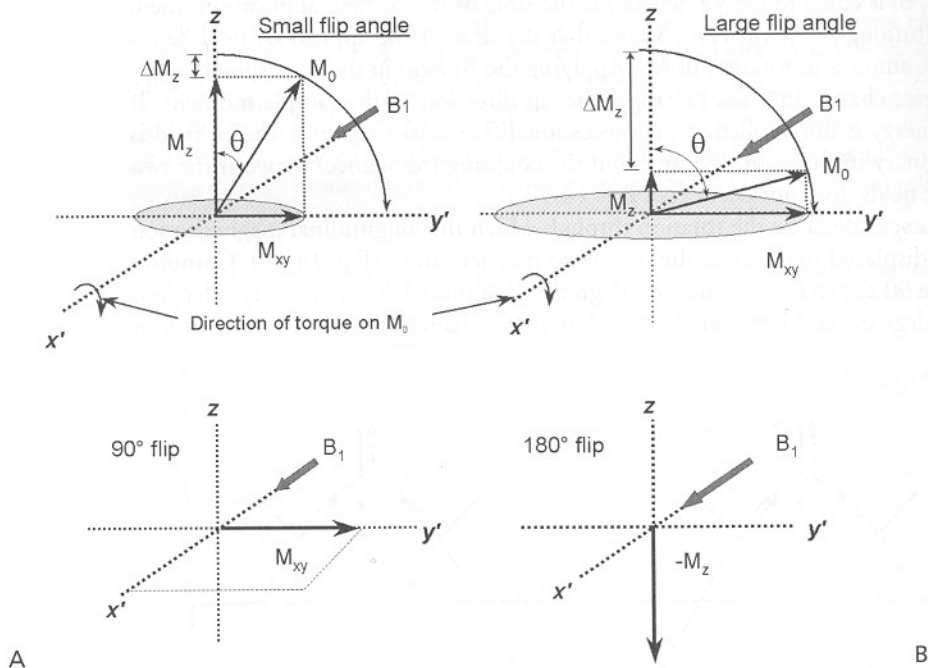


FIGURE 14-10. A: Flip angles describe the angular displacement of the longitudinal magnetization vector from the equilibrium position. The rotation angle of the magnetic moment vector depends on the duration and amplitude of the B_1 field at the Larmor frequency. Small flip angles (~ 30 degrees) and large flip angles (~ 90 degrees) produce small and large transverse magnetization, respectively. **B:** Common flip angles are 90 degrees, which produces the maximum transverse magnetization, and 180 degrees, which inverts the existing longitudinal magnetization M_z to $-M_z$.

ways. A 90-degree angle provides the largest possible transverse magnetization. The time required to flip the magnetic moment is linearly related to the displacement angle: For the same B_1 field strength, a 90-degree angle takes half the time to produce that a 180-degree angle does. The time required to implement a rotation is on the order of tens to hundreds of microseconds. With fast MR imaging techniques, 30-degree and smaller angles are often used to reduce the time needed to displace the longitudinal magnetization and generate the transverse magnetization. For flip angles smaller than 90 degrees, less signal in the M_{xy} direction is generated, but less time is needed to displace M_z , resulting in a *greater amount of transverse magnetization (signal) per excitation time*. For instance, a 45-degree flip takes half the time of a 90-degree flip yet creates 70% of the signal, because the projection of the vector onto the transverse plane is $\sin 45$ degrees, or 0.707. In instances where short excitation times are necessary, small flip angles are employed.

Free Induction Decay: T2 Relaxation

The 90-degree RF pulse produces phase coherence of the individual protons and generates the maximum possible transverse magnetization for a given sample volume. As M_{xy} rotates at the Larmor frequency, the receiver antenna coil (in the *laboratory* frame) is induced (by *magnetic induction*) to produce a damped sinusoidal electronic signal known as the *free induction decay* (FID) signal, as shown in Fig. 14-11.

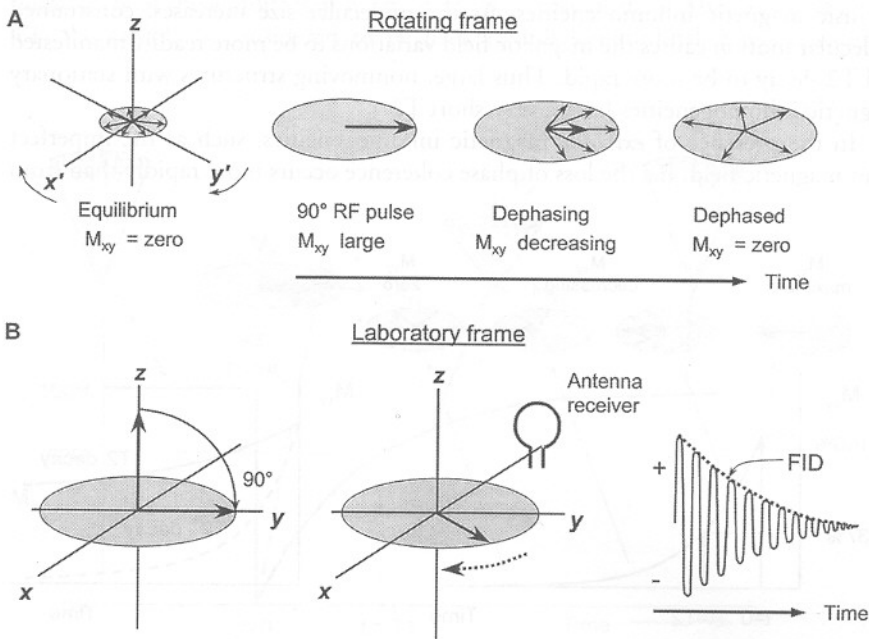


FIGURE 14-11. A: Conversion of longitudinal magnetization, M_z , into transverse magnetization, M_{xy} , results in an initial phase coherence of the individual spins of the sample. The magnetic moment vector precesses at the Larmor frequency (stationary in the rotating frame) and dephases with time. **B:** In the laboratory frame, M_{xy} precesses and induces a signal in an antenna receiver that is sensitive to transverse magnetization. A free induction decay (FID) signal is produced, oscillating at the Larmor frequency, and decays with time due to the loss of phase coherence.

The "decay" of the FID envelope is the result of the loss of phase coherence of the individual spins caused by magnetic field variations. Micromagnetic inhomogeneities intrinsic to the structure of the sample cause a spin-spin interaction, whereby the individual spins precess at different frequencies due to slight changes in the local magnetic field strength. Some spins travel faster and some slower, resulting in a loss of phase coherence. External magnetic field inhomogeneities arising from imperfections in the magnet or disruptions in the field by paramagnetic or ferromagnetic materials accelerate the dephasing process. Exponential relaxation decay, T_2 , represents the intrinsic *spin-spin interactions* that cause loss of phase coherence due to the intrinsic magnetic properties of the sample. The elapsed time between the peak transverse signal and 37% of the peak level ($1/e$) is the T_2 decay constant (Fig. 14-12A). Mathematically, this exponential relationship is expressed as follows:

$$M_{xy}(t) = M_0 e^{-t/T_2}$$

where M_{xy} is the transverse magnetic moment at time t for a sample that has M_0 transverse magnetization at $t = 0$. When $t = T_2$, then $e^{-1} = 0.37$, and $M_{xy} = 0.37 M_0$. An analogous comparison to T_2 decay is that of radioactive decay, with the exception that T_2 is based on $1/e$ decay instead of half-life ($1/2$) decay. This means that the time for the FID to reach half of its original intensity is given by $t = 0.693 \times T_2$.

T_2 decay mechanisms are determined by the molecular structure of the sample. Mobile molecules in amorphous liquids (e.g., cerebral spinal fluid [CSF]) exhibit a long T_2 , because fast and rapid molecular motion reduces or cancels intrinsic magnetic inhomogeneities. As the molecular size increases, constrained molecular motion causes the magnetic field variations to be more readily manifested and T_2 decay to be more rapid. Thus large, nonmoving structures with stationary magnetic inhomogeneities have a very short T_2 .

In the presence of *extrinsic* magnetic inhomogeneities, such as the imperfect main magnetic field, B_0 , the loss of phase coherence occurs more rapidly than from

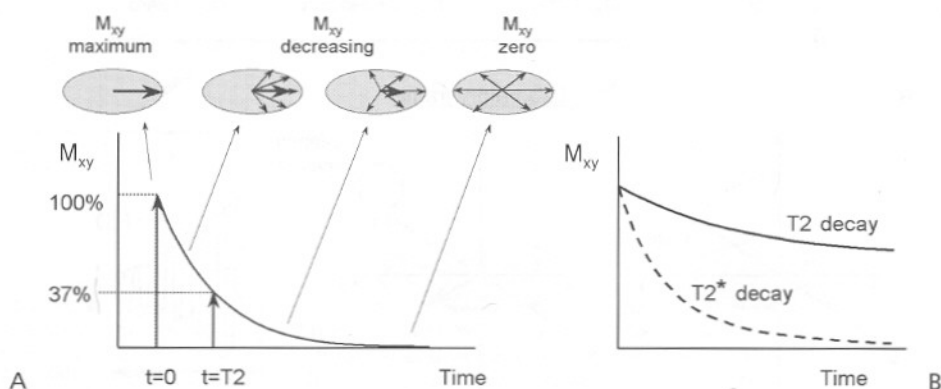


FIGURE 14-12. A: The loss of M_{xy} phase coherence occurs exponentially and is caused by intrinsic spin-spin interactions in the tissues, as well as extrinsic magnetic field inhomogeneities. The exponential decay constant, T_2 , is the time over which the signal decays to 37% of the maximal transverse magnetization (e.g., after a 90-degree pulse). **B:** T_2 is the decay time that results from intrinsic magnetic properties of the sample. T_2^* is the decay time resulting from both intrinsic and extrinsic magnetic field variations. T_2 is always longer than T_2^* .

spin-spin interactions by themselves. When B_0 inhomogeneity is considered, the spin-spin decay constant T_2 is shortened to T_2^* . Figure 14-12B shows a comparison of the T_2 and T_2^* decay curves. T_2^* depends on the homogeneity of the main magnetic field and susceptibility agents that are present in the tissues (e.g., MR contrast materials, paramagnetic or ferromagnetic objects).

Return to Equilibrium: T1 Relaxation

The loss of transverse magnetization (T_2 decay) occurs relatively quickly, whereas the return of the excited magnetization to equilibrium (maximum longitudinal magnetization) takes a longer time. Individual excited spins must release their energy to the local tissue (the lattice). *Spin-lattice relaxation* is a term given for the exponential *regrowth* of M_z , and it depends on the characteristics of the spin interaction with the *lattice* (the molecular arrangement and structure). The T_1 relaxation constant is the time needed to *recover* 63% of the longitudinal magnetization, M_z , after a 90-degree pulse (when $M_z = 0$). The recovery of M_z versus time after the 90-degree RF pulse is expressed mathematically as follows:

$$M_z(t) = M_0(1 - e^{-t/T_1})$$

where M_z is the longitudinal magnetization that recovers after a time t in a material with a relaxation constant T_1 . Figure 14-13 illustrates the recovery of M_z . When $t = T_1$, then $1 - e^{-1} = 0.63$, and $M_z = 0.63 M_0$. Full longitudinal recovery depends on the T_1 time constant. For instance, at a time equal to $3 \times T_1$ after a 90-degree pulse, 95% of the equilibrium magnetization is reestablished. After a period of $5 \times T_1$, the sample is considered to be back to full longitudinal magnetization. A

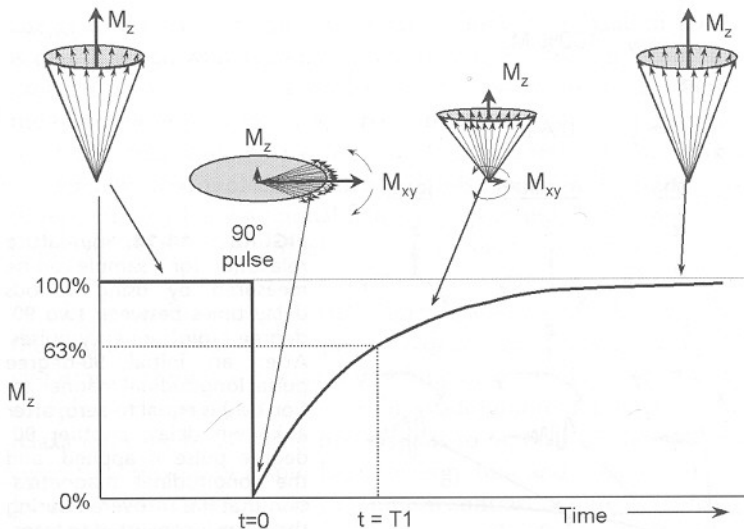


FIGURE 14-13. After a 90-degree pulse, longitudinal magnetization (M_z) is converted from a maximum value at equilibrium to zero. Return of M_z to equilibrium occurs exponentially and is characterized by the spin-lattice T_1 relaxation constant. After an elapsed time equal to T_1 , 63% of the longitudinal magnetization is recovered. Spin-lattice recovery takes longer than spin-spin decay (T_2).

method to determine the T_1 time of a specific tissue or material is illustrated in Fig. 14-14. An initial 90-degree pulse, which takes the longitudinal magnetization to zero, is followed by a delay time, ΔT , and then a second 90-degree pulse is applied to examine the M_z recovery by displacement into the transverse plane (only M_{xy} magnetization can be directly measured). By repeating the sequence with different delay times, ΔT , between 90-degree pulses (from equilibrium conditions), data points that lie on the T_1 recovery curve are determined. The T_1 value can be estimated from these values.

T_1 relaxation depends on the dissipation of absorbed energy into the surrounding molecular lattice. The relaxation time varies substantially for different tissue structures and pathologies. From a classical physics perspective, energy transfer is most efficient when the precessional frequency of the excited protons overlaps with the "vibrational" frequencies of the molecular lattice. Large, slowly moving molecules exhibit low vibrational frequencies that concentrate in the lowest part of the frequency spectrum. Moderately sized molecules (e.g., proteins) and viscous fluids produce vibrations across an intermediate frequency range. Small molecules have vibrational frequencies with low-, intermediate-, and high-frequency components that span the widest frequency range. Therefore, T_1 relaxation is strongly dependent on the physical characteristics of the tissues, as illustrated in Fig. 14-15. Consequently, for *solid and slowly moving structures*, low-frequency variations exist and there is little spectral overlap with the Larmor frequency. A small spectral overlap also occurs for unstructured tissues and fluids that exhibit a wide vibrational frequency spectrum but with low amplitude. In either situation, the inability to release

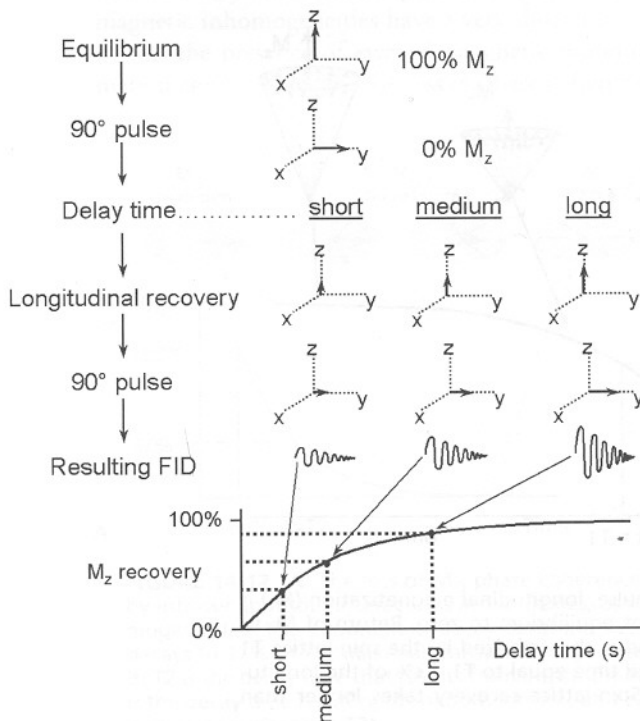


FIGURE 14-14. Spin-lattice relaxation for a sample can be measured by using various delay times between two 90-degree radiofrequency pulses. After an initial 90-degree pulse, longitudinal magnetization (M_z) is equal to zero; after a known delay, another 90-degree pulse is applied, and the longitudinal magnetization that has recovered during the delay is converted to transverse magnetization. The maximum amplitude of the resultant free induction decay (FID) is recorded as a function of delay time, and the points are fit to an exponential recovery function to determine T_1 .

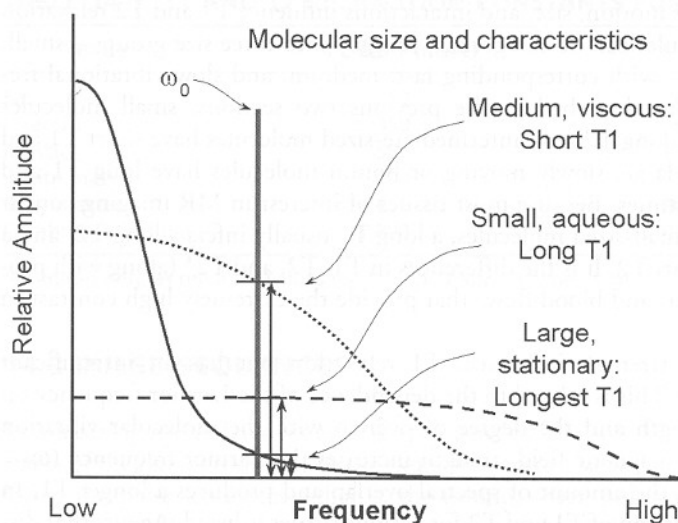


FIGURE 14-15. A classical physics explanation of spin-lattice relaxation is based on the vibration of the molecular lattice of a sample material and its frequency spectrum. Large, stationary structures exhibit little motion with mostly low-frequency vibrations (**solid curve**). Medium sized, proteinated materials have increased frequency amplitudes (**short dash curve**), and small sized, aqueous materials have frequencies distributed over a broad range (**long dash curve**). The overlap of the Larmor precessional frequency (**vertical bar**) with the molecular vibration spectrum indicates the probability of spin-lattice relaxation.

energy to the lattice results in a relatively long T_1 relaxation. One interesting case is that of water, which has an extremely long T_1 , but the addition of water-soluble proteins produces a hydration layer that slows the molecular vibrations and shifts the high frequencies in the spectrum to lower values that increase the amount of spectral overlap with the Larmor frequency and result in a dramatically shorter T_1 . Moderately sized molecules, such as lipids, proteins, and fats, have a more structured lattice with a vibrational frequency spectrum that is most conducive to spin-lattice relaxation. For biologic tissues, T_1 ranges from 0.1 to 1 second in soft tissues, and from 1 to 4 seconds in aqueous tissues (e.g., CSF) and water.

T_1 relaxation increases with higher field strengths. A corresponding increase in the Larmor precessional frequency reduces the spectral overlap of the molecular vibrational frequency spectrum, resulting in longer T_1 times. Contrast agents (e.g., complex macromolecules containing gadolinium) are effective in decreasing T_1 relaxation time by allowing free protons to become bound and create a hydration layer, thus providing a spin-lattice energy sink and a rapid return to equilibrium. Even a very small amount of gadolinium contrast in pure water has a dramatic effect on T_1 , decreasing the relaxation from a couple of seconds to tens of milliseconds!

Comparison of T_1 and T_2

T_1 is significantly longer than T_2 . For instance, in a soft tissue, a T_1 time of 500 msec has a corresponding T_2 time that is typically 5 to 10 times shorter (i.e., about

50 msec). Molecular motion, size, and interactions influence T1 and T2 relaxation (Fig. 14-16). Molecules can be categorized roughly into three size groups—small, medium, and large—with corresponding fast, medium, and slow vibrational frequencies. For reasons described in the previous two sections, small molecules exhibit long T1 and long T2, and intermediate-sized molecules have short T1 and short T2; however, large, slowly moving or bound molecules have long T1 and short T2 relaxation times. Because most tissues of interest in MR imaging consist of intermediate to small-sized molecules, a long T1 usually infers a long T2, and a short T1 infers a short T2. It is the differences in T1, T2, and T2* (along with proton density variations and blood flow) that provide the extremely high contrast in MRI.

Magnetic field strength influences T1 relaxation but has an insignificant impact on T2 decay. This is related to the dependence of the Larmor frequency on magnetic field strength and the degree of overlap with the molecular vibration spectrum. A higher magnetic field strength increases the Larmor frequency ($\omega_0 = \gamma B_0$), which reduces the amount of spectral overlap and produces a longer T1. In Table 14-4, a comparison of T1 and T2 for various tissues is listed. Agents that disrupt the local magnetic field environment, such as paramagnetic blood degradation products, elements with unpaired electron spins (e.g., gadolinium), or any ferromagnetic materials, cause a significant decrease in T2*. In situations where a macromolecule binds free water into a hydration layer, T1 is also significantly decreased.

To summarize, $T1 > T2 > T2^*$, and the specific relaxation times are a function of the tissue characteristics. The spin density, T1, and T2 decay constants are fundamental properties of tissues, and therefore these tissue properties can be exploited by MRI to aid in the diagnosis of pathologic conditions such as cancer, multiple sclerosis, or hematoma. It is important to keep in mind that T1 and T2 (along with spin density) are fundamental properties of tissue, whereas the other time-dependent parameters are machine-dependent. Mechanisms to exploit the differences in T1, T2, and spin density (proton density) are detailed in the next section.

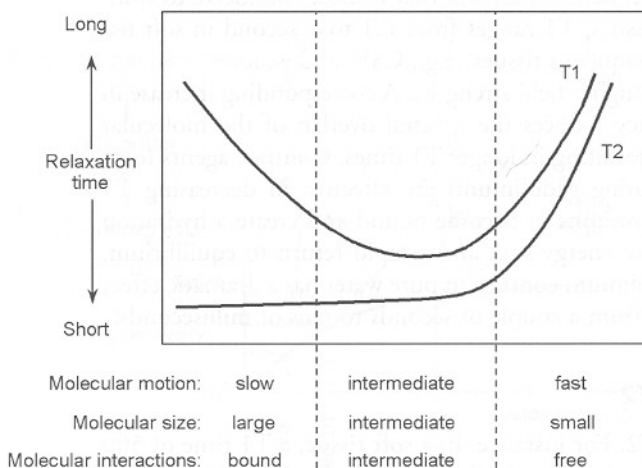


FIGURE 14-16. Factors affecting T1 and T2 relaxation times of different tissues are generally based on molecular motion, size, and interactions. The relaxation times (vertical axis) are different for T1 and T2.