MRI for Technologists

Basic Principles of MRI

PROGRAM INFORMATION

MRI for Technologists is a training program designed to meet the needs of radiologic technologists entering or working in the field of magnetic resonance imaging (MRI). These units are designed to augment classroom instruction and on-site training for radiologic technology students and professionals planning to take the review board examinations, as well as to provide a review for those looking to refresh their knowledge base in MR imaging.

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This material will be reviewed for continued accuracy and relevance. Please go to www.icpme.us for up-to-date information regarding current expiration dates.

OVERVIEW

The skill of the technologist is the single most important factor in obtaining good quality diagnostic images. A successful MRI examination is the culmination of many factors under the direct control of the technologist.

Basic Principles of MRI introduces the learner to the fundamental technical concepts of magnetic resonance imaging including the physics of how hydrogen protons respond when subjected to a magnetic field to how changes in magnetization can be detected and recorded.

After completing this educational material, the reader will be able to:

- List the different types of tomographic imaging
- Explain how MRI and CT differ
- Explain the atomic structure of the hydrogen proton and its utility in MRI
- Describe how the hydrogen proton responds when placed in an external magnetic field
- Explain how a rotating magnetic field affects the behavior of hydrogen protons
- Describe how protons can be reoriented to longitudinal and transverse directions
- Compare and contrast transverse and longitudinal relaxation
- Describe the time constants relevant to transverse and longitudinal relaxation
- Explain Faraday's law
- Discuss why and when free induction decay occurs

EDUCATIONAL CREDIT

This program has been approved by the American Society of Radiologic Technologists (ASRT) for 2.0 hours ARRT Category A continuing education credit.

HOW TO RECEIVE CREDIT

Estimated time to complete this activity is 2.0 hours. The posttest and evaluation are required to receive credit and must be completed online.

- In order to access the posttest and evaluation, enroll in the online course at icpme.us.
- Read the entire activity.
- Log in to your account at icpme.us to complete the posttest and evaluation, accessible through the course link in your account.
- A passing grade of at least 75% is required to be eligible to receive credit.
- You may take the test up to three times.
- Upon receipt of a passing grade, you will be able to print a credit certificate of credit from your online account.

FACULTY

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Dr. Theden's research interests are 3D MR image acquisition, rapid MR acquisition techniques, imaging of cartilage and other orthopaedic applications, cardiac MRI, and MR image processing.

We are grateful to Dr. Thedens for updating his original work, released in 2009.

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Basic Principles of MRI

*Please note: items in **bold** can be found in the glossary.*

After completing this material, the reader should be able to:

- List the different types of tomographic imaging
- Explain how MRI and CT differ
- Explain the atomic structure of the hydrogen proton and its utility in MRI
- Describe how the hydrogen proton responds when placed in an external magnetic field
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- Describe how protons can be reoriented to longitudinal and transverse directions
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INTRODUCTION and OVERVIEW

Magnetic resonance imaging (MRI) has become one of the most powerful and versatile diagnostic tools available to clinicians and researchers today. There has been tremendous growth in both technology and clinical usage since MRI first became commercially available.

The resolution, image quality, and time required for MRI exams have improved significantly with the design of robust high-field magnet designs, high-performance gradient systems, a wide range of specialized radiofrequency coils, and advancement of computer technologies. These new capabilities have yielded exquisitely detailed images in all areas of the body, bringing advanced technologies such as functional brain imaging, whole-body diffusion imaging, fiber tracking, and real-time cardiac imaging out of the research lab and into clinical practice.

Improvements in hardware and software technology have also permitted the development of user-friendly, sophisticated viewing and processing tools to further enhance the diagnostic ability of clinicians. As a result, MRI is now routinely used for an ever-expanding range of diagnostic examinations, providing better information to clinicians, reducing risk to patients by eliminating the need for radiation and invasive procedures, and yielding improved diagnosis and treatment plans.

Development of MRI:
Discovery, Invention, and Nobel Prizes

MRI relies on the physical principles of nuclear magnetic resonance (NMR). In 1945 in independent experiments, Felix Bloch at Stanford University and Edward Purcell at Harvard University discovered that sending certain radio waves into materials subjected to a strong magnetic field caused the material to absorb the energy of the waves which could then be detected as the energy radiated back.

POINTS for PRACTICE

1. In addition to MRI, what are other primary types of tomographic imaging?

2. One of your friends is familiar with CT and wants to know how MRI differs from CT. How would you explain this? Overall, what are some of the technical and clinical advantages of using MRI?

Further experiments found that the precise set of absorbed frequencies provided information about the structure of the atomic nucleus and its chemical environment. By studying the spectrum of absorbed frequencies, the structure of complex molecules could be determined. This analysis technique is known as nuclear magnetic resonance (NMR) spectroscopy or magnetic resonance spectroscopy (MRS) and continues to be a primary means for identifying the structure of proteins and many other molecules. In 1952, Bloch and Purcell were awarded the Nobel Prize in Physics for their pioneering work in nuclear magnetic resonance experimentation.

Technical Milestones

Recognizing the ability of NMR to identify molecular changes in tissues, Raymond Damadian proposed using NMR to discriminate between healthy and cancerous tissues. In 1971, he was able to demonstrate differences between NMR properties of normal and abnormal tissues and tumors in rats. Two years later, Paul Lauterbur showed that adding an additional magnetic field with strength dependent on the location within the sample (a gradient field) made it possible to map both the location and the distribution of the tissue in the field. Consequently, he generated the very first magnetic resonance image of a pair of water-filled tubes, and the field of magnetic resonance imaging was born. As other investigators continued to modify and adapt NMR techniques for imaging, the word “nuclear” was dropped, and the modality came to be called MRI.

Research in MRI advanced rapidly. The 1970s saw the development of many of the fundamental techniques still used in MRI today: slice selection, phase encoding, and echo-planar imaging. By 1977, Peter Mansfield, another pioneer in MRI, obtained images of a human finger and later an abdominal cross-section. Damadian was able to acquire whole body images, and in 1980 his company, FONAR, produced the first commercially available scanner. By 1984, several companies had received FDA approval for MRI magnets for clinical use, and within a year more than 90 scanners had been installed in the United States. Lauterbur and Mansfield were awarded the 2003 Nobel Prize for Physiology or Medicine for their discoveries and developments.

Innovations in MRI continued at a rapid pace. Magnetic resonance angiography (MRA), capable of imaging the blood vessels throughout the body, appeared in 1986. In 1988, gadopentetate dimeglumine became the first FDA-approved contrast agent for MRI**.** Functional MRI (fMRI), first developed in 1993, probes the areas of the brain activated by an extremely wide range of tasks and is emerging as a clinical tool for surgical planning. The introduction of **parallel imaging** coils in the late 1990s and processing techniques have yielded remarkable reductions in scan time. Continued advances in computer processing power and acquisition techniques, such as compressed sensing, are poised to produce even faster and more detailed images. 3T and higher field magnets are also showing never-before-seen anatomical and functional details. There is reason to believe that development of MRI will continue to progress into the foreseeable future.

Figure 1. Short bore MRI.

Courtesy of Siemens Medical Systems.

In addition to these technical advances, MRI scanners have evolved to be easier to install and more patient-friendly and user-friendly. Compared to earlier scanners, today's scanners have a relatively short bore and compact design that help reduce patient anxiety and claustrophobia (**Figure 1**).

MRI AND OTHER IMAGING MODALITIES

X-ray

In addition to MRI, there is a wide range of imaging techniques routinely used in the clinical setting. X-ray imaging, also called radiography, was the first, and for decades the only, available imaging method able to "see into the body." X-rays are a type of electromagnetic radiation that exposes tissues to ionizing radiation which accumulates in the body over the patient's lifetime. It remains among the most widely performed exams in medicine. X-ray imaging is quick, inexpensive, and can even be portable.

There are several types of specialized x-ray techniques. 2D mammography utilizes x-ray technology and is the gold standard for breast imaging. 3D mammography (digital breast tomosynthesis), approved by the FDA in 2012, is rapidly gaining acceptance as an alternative for breast cancer screening. Fluoroscopy is an x-ray technique that produces images in real-time. Each of these specialized x-ray techniques depicts the absorption of x-rays as they pass through the body tissues, which may absorb x-rays differently.

Because x-rays can only be taken as projections through the body, limiting the views of internal organs, they may not provide the necessary information needed for diagnosis. What follows is a discussion of several other imaging techniques that have been developed in the past decades to address the gaps and limitations of x-ray imaging.

Tomographic Imaging

MRI represents one of several noninvasive techniques that produce cross-sectional images — known as **tomographic** images — available to radiologists and referring clinicians. Other widely used tomographic modalities include computed tomography (CT), nuclear medicine, single photon emission computed tomography (SPECT), positron emission tomography (PET), and ultrasound (US).

Computed Tomography

Among the tomographic modalities, CT is most commonly compared to MRI because each modality yields images of anatomy at broadly similar levels of detail with differing shades of gray, and both provide images from anywhere in the body. However, the principles of how images are formed and the appearance of specific tissues in each modality are very different.

As with MRI and all other imaging methods, CT forms an image by sending energy into the body and measuring how that energy is absorbed or changed when passing through the body's tissues. In CT, that energy is in the form of x-ray beams passed through the body. Detectors measure how much energy passes through the body on the opposite side, which in turn tells how much the beam was attenuated, that is, how much energy was absorbed by the tissues. The x-ray beam is applied at many different angles, and the resulting measurements are processed by a computer to produce a cross-sectional image.

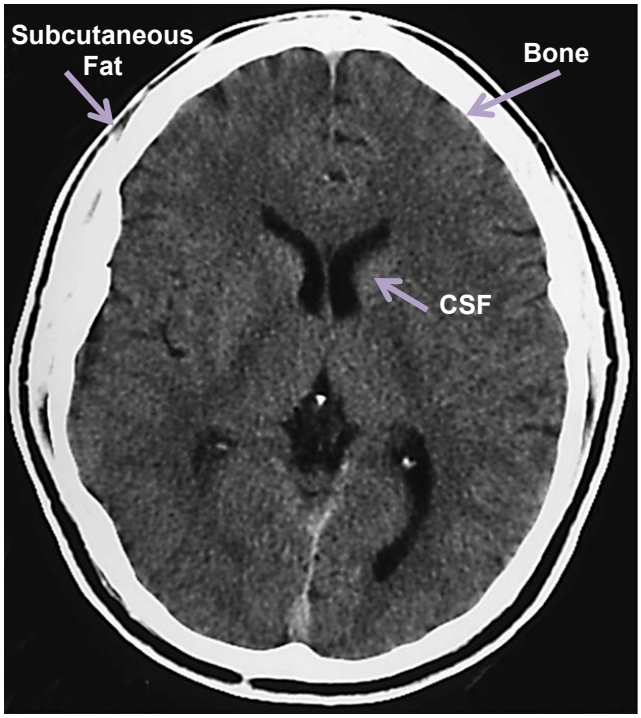

Figure 2. Axial CT image of the brain.

Courtesy of Maimonides Medical Center, Brooklyn, NY.

Tissue Characterization on CT

Both the displayed brightness of the tissue at each point and the contrast between different tissues in the image reflect how much x-ray energy is absorbed. For example, bone is one of the more dense tissues in the body, so it absorbs more x-rays and thus appears bright on a CT image. Blood, cerebrospinal fluid (CSF), and some soft tissues are of lower density so absorb much less of the x-ray beam, resulting in a darker appearance on the image (**Figure 2**).

Tissue Characterization on MRI

Both image brightness and contrast in MRI are based on completely different principles than for CT. In MRI, brightness depicts the differences in how tissues interact with radio waves, not x-ray, inside a magnetic field. The resulting interactions cause the tissues to appear bright, dark, or a shade of gray somewhere in-between. This modality, then, reflects differences in the molecular composition and environment of the tissues. In particular, diverse soft tissues will demonstrate differences in their MRI appearance. Additionally, the radio waves and magnetic fields used in MRI can be manipulated in many ways to alter the appearance of specific tissues for greater versatility. Tissues that consist mostly of water, like CSF, can be made to appear either bright or dark on MRI. For example, on T1-weighted MRI brain images, fluids appear dark; on T2-weighted MRI images, fluids appear bright (**Figure 3**). MRI acquisitions can also be made to show differences in blood flow, fat content, diffusion characteristics, and a host of other traits.

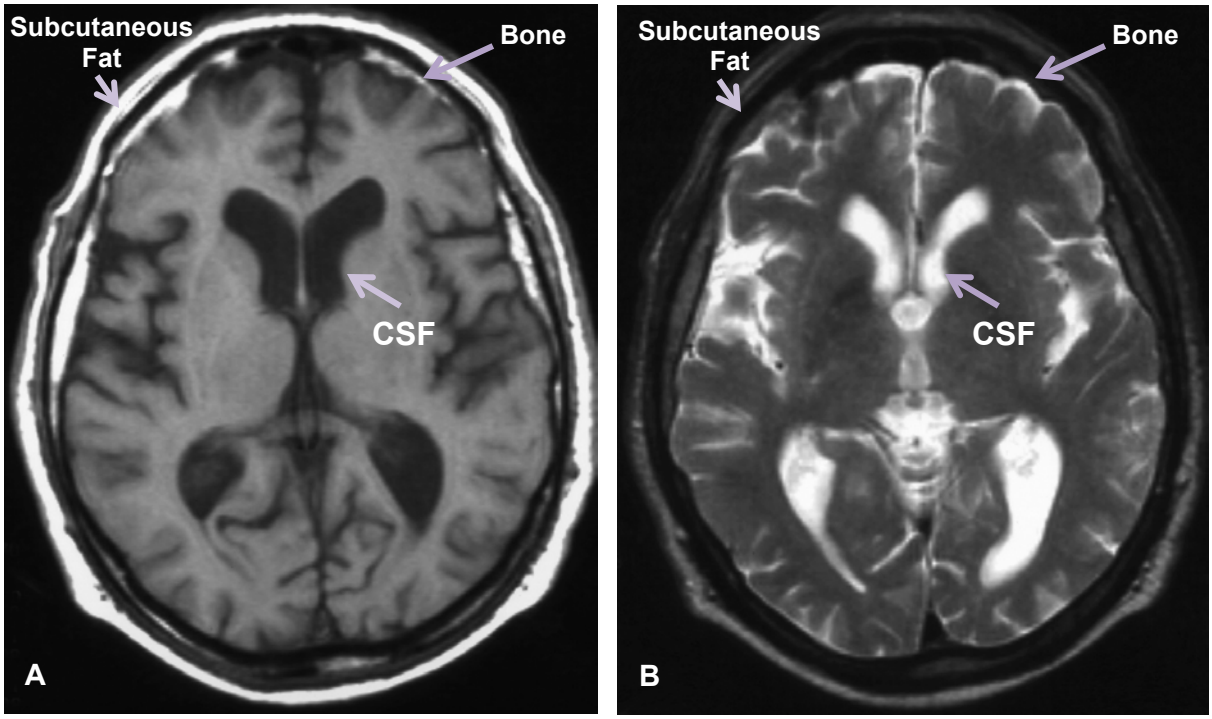

Figure 3. MRI of the brain. (A) T1-weighted axial image. (B) T2-weighted axial image.

Courtesy of Maimonides Medical Center, Brooklyn, NY.

MRI vs CT in Clinical Practice

Both MRI and CT are widely used for diagnosis in routine clinical practice. In some instances, MRI may be chosen over CT because:

- MRI uses no ionizing radiation; there is little risk of tissue damage from repeated scans.
- MRI acquires images directly in any orientation.
- MRI better differentiates contrast between different kinds of soft tissue.
- MRI generates images with different tissue contrast properties, with or without the use of contrast agent injection.
- MR angiography is capable of directly measuring and quantifying the direction and velocity of blood flow.
- MRI contrast media are generally better tolerated than CT contrast media.

Nuclear Medicine, SPECT, PET, and US

Both nuclear medicine and SPECT work by introducing radioactive energy via a radioactive tracer into the body and measuring the decay of that tracer to form an image of the radioactive quantity and distribution. Injected contrast agents emit gamma rays as they decay, similar to x-rays but with higher energy.

Gamma rays are detected by a gamma camera that forms the final image. Most commonly, nuclear medicine and SPECT are used to measure the flow and distribution of blood into the heart and other organs. Compared to MRI, these modalities produce images of much lower resolution but are specifically tailored to measuring organ function.

PET relies on a radioactive contrast agent as well as detectors that measure its distribution from the radioactive decay. In PET, however, the decay produces tiny particles called positrons that are detected by the scanning device. PET imaging is unique because the radioactive material used can be attached to various substances like glucose. The amount of glucose accumulated in a tissue relates to tissue metabolism, and thus normal tissue can be differentiated from cancerous tissue. PET images are of lower resolution compared to MRI but provide unique information on organ function. There are also hybrid systems, for example PET/CT and PET/MRI, that combine both a high resolution anatomical image and a corresponding image of metabolism in the same exam. Areas of metabolic activity can be directly correlated to anatomic location, which is useful in tumor ablation.

Among the other tomographic imaging methods, ultrasound is the most widely available. It is comparatively inexpensive, and the machines are small and portable. Ultrasound is routinely used in fetal assessment during pregnancy and for scanning the heart, major arteries, liver, and kidneys. Like MRI, no ionizing radiation is used; instead, ultrasound measures the energy of sound waves. Because sound does not travel well through air or bone, some areas of the body, such as the lungs and the skeleton, cannot be optimally imaged using ultrasound.

Limitations of MRI

MRI provides unique information about the anatomy and function of tissues throughout the body. Nevertheless, there are circumstances where MRI is of limited use or may not be desirable. Because of safety risks, MRI cannot be used in patients with **ferromagnetic** metal in the body as a result of an accident, occupational hazards, or surgical implantation. Most modern metal implants, eg, stents and orthopaedic screws, are made of MRI-compatible materials and are safe to scan. However, if metal implants are in or near the area of interest, they may generate significant artifact that make the scan unreadable. This phenomenon will worsen at higher field strengths.

Another limiting factor is the typical length of an MRI exam, which is relatively longer than that of CT or US, and patient movement during this time may cause image artifacts. In addition, patients who become claustrophobic in the closed magnet may have difficulty tolerating the exam. In such cases, other imaging methods may be indicated (**Table 1**).

Despite its limitations, MRI is currently one of the most widely available and routinely used examinations for assessing tissues and pathology throughout the body. Compared to CT and other imaging methods, MRI provides unsurpassed versatility in its range of applications and clinical uses.

MODALITY	APPROXIMATE RESOLUTION	ENERGY SOURCE	SOURCE OF IMAGE CONTRAST	ADVANTAGES	LIMITATIONS
CT	0.2-1.0mm	x-ray	tissue density	fast, high resolution, 3D reconstruction	poor soft tissue contrast, radiation dose
MRI	0.3-1.0mm	radiofrequency	multiple	soft tissue contrast, multiple contrast methods	scan times, metal, patient comfort
NUC MED	5-10mm	gamma rays	tissue biochemistry	tissue function information	low resolution, radioactive elements
PET	4-7mm	positrons/ gamma rays	tissue biochemistry	tissue function information	low resolution, radioactive elements
SPECT	5-10mm	gamma rays	tissue biochemistry	tissue function information	low resolution, radioactive elements
US	0.2-0.5mm	acoustic (sound)	tissue composition or flow	fast, inexpensive, portable, real-time	poor image quality, limited field of view
X-RAY	0.03-0.2mm	x-ray	tissue density	fast, inexpensive, high resolution	2D only, poor soft tissue contrast

Table 1. Comparison of various imaging modalities.

POINTS for PRACTICE

1. In addition to MRI, what are other primary types of tomographic imaging?

- computed tomography (CT)
- nuclear medicine
- positron emission tomography (PET)
- single photon emission computed tomography (SPECT)
- ultrasound (US)

2. One of your friends is familiar with CT and wants to know how MRI differs from CT. How would you explain this? Overall, what are some of the technical and clinical advantages of using MRI?

MRI is based on the nuclear magnetic properties of atoms, while CT is based on the attenuation of x-rays. Both are used to acquire cross-sectional images and are versatile and prominent diagnostic imaging modalities.

MRI uses radio waves – not x-ray – to acquire images, with little risk of tissue damage. MRI better differentiates contrast between soft tissues and can produce direct multiplanar images. It can also measure and quantify blood flow. MRI is not associated with ionizing radiation, and MR contrast media are generally better tolerated than those used in CT.

NOTES

DESCRIPTION and CHARACTERISTICS OF ATOMS

Magnetic resonance imaging is based on the principle of nuclear magnetic resonance, where "nuclear" refers to the nucleus of atoms. To understand how the signal used to form a magnetic resonance image is produced, we need to understand the structure of atoms.

Atomic Structure

An atom is the smallest unit of a chemical element and is made up of three types of particles: protons, neutrons, and electrons. The nucleus is the dense core of the atom and contains protons and neutrons. The chemical identity of an atom is determined by the number of protons contained in its nucleus; the number of protons is known as its atomic number. All atoms of any given element contain the same number of protons in their nucleus and therefore have the same atomic number. Protons also carry a positive electrical charge of one unit. The number of protons in the nucleus determines many of its chemical properties.

POINTS for PRACTICE

1. What particles make up an atom? How does an atom become ionized?

2. An electrically neutral atom has eleven particles in its nucleus and five particles orbiting the nucleus. How many protons does it have? neutrons? electrons?

3. What chemical element is most often used for MRI imaging and why? What other elements can be used in MRI?

4. Why can a positively charged hydrogen atom be described as a spinning top?

The nuclei of most atoms also contain neutrons. Neutrons are particles that are almost the same size as protons but do not carry any electrical charge — they are neutral. Since the nucleus is made up of positively charged protons and neutral neutrons, the nucleus of any atom has an overall positive electrical charge.

The nucleus of the atom is surrounded by a cloud of electrons that moves rapidly and orbits around the nucleus (**Figure 4**). Electrons are much smaller than protons and neutrons and have a negative electrical charge of one unit. In a neutral atom, the number of electrons surrounding the nucleus is equal to the number of protons, that is, the positive charge of the nucleus created by the protons is exactly balanced by the negative charge of the electrons. In certain circumstances, there may be electrons gained or lost in the orbits around the nucleus, leaving the atom with a net positive or net negative charge. When this occurs, the atom is said to be **ionized**.

Magnetic Fields

In addition to the electrical properties of atoms, the particular arrangement of particles in an atom affects their behavior in a magnetic field. To better appreciate this behavior, we need to understand some of the properties of magnets and magnetic fields.

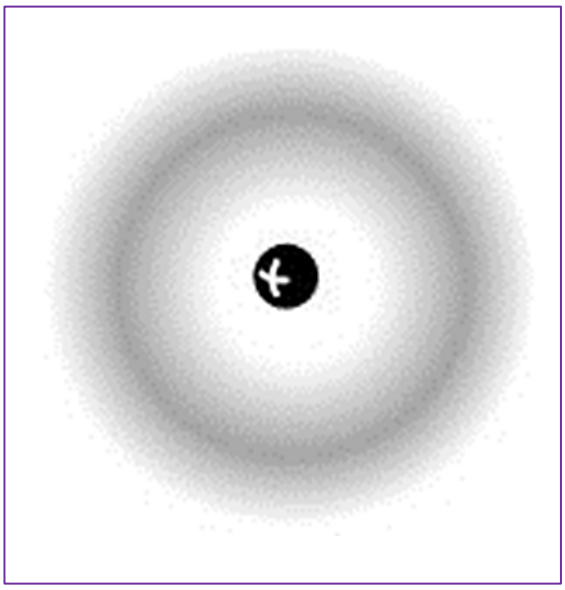

Figure 4. Illustration of a hydrogen atom containing a positively charged proton in the nucleus and a negatively charged electron orbiting in a 'cloud' surrounding the nucleus.

Courtesy of JD Norton, University of Pittsburgh Center for Philosophy and Science. Available at: University of Pittsburgh.

A magnet is an object that attracts iron as well as a few other substances. Magnets come in all shapes and sizes and are common in everyday applications. Small bar magnets are often found on refrigerators to post notes. Most audio speakers contain larger magnets. A typical compass contains a magnetized pointer that works because the earth itself is a magnet, with magnetic north and south poles. The poles of the compass needle line up with the earth's magnetic field to indicate north and south. An object may naturally be a magnet or it may be induced by magnetizing the object.

A magnet has two distinct ends called poles. They are referred to as the north and south poles, and the magnet with these two distinct ends is called a **dipole**. When two magnets are brought together, the north pole of one magnet is attracted to the south pole of the other. When two north or south poles are brought together, they will repel each other.

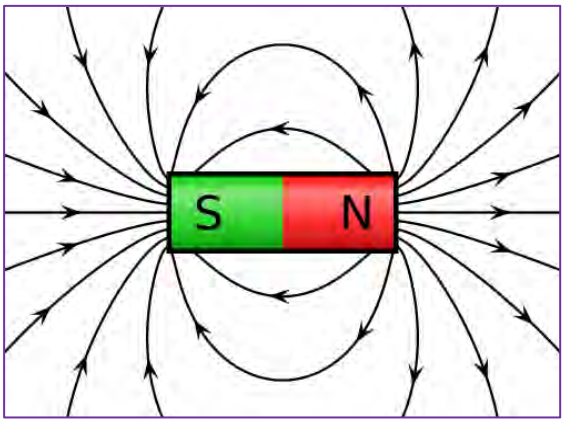

Figure 5. All magnets have a north pole and a south pole and are surrounded by a magnetic field.

Available at Wikimedia.

Surrounding the magnet is a magnetic field that describes the strength **(magnitude)** and direction of the forces created by the magnet (**Figure 5**).

Because this magnetic force is characterized by both a direction and a strength, to describe it fully we use a **vector** to symbolize these quantities. A vector is usually drawn as an arrow pointing along the orientation of the magnetic field, while the length of the drawn arrow is proportional to the strength of the field.

Magnetic Properties

Elements that are ferromagnetic - most commonly iron - are strongly attracted by a magnetic field. Ferromagnetic elements retain their magnetic properties after the magnetic field is removed, acting as magnets themselves. An element that is **paramagnetic** is *slightly* attracted to a magnetic field. Such elements have unpaired electron spins that produce a small amount of magnetism or **magnetic moments**. Examples of paramagnetic elements include oxygen and gadolinium. An element is **diamagnetic** if the element is *slightly* repelled by a magnetic field. Helium, copper, and gold are examples of diamagnetic elements. The strength of the attraction or repulsion of paramagnetic and diamagnetic elements is very weak and usually not discernible by ordinary bar magnets.

Characteristics of Atoms Used in MRI

A number of elements can be used to form images with MRI. Logically, we would like an element that is abundant in the molecules of body tissues; recall that a molecule is a collection of atoms of the same or different elements bound together. By far the most common element that can be used for MRI in body tissues is hydrogen because it is abundant in water, protein, and fat molecules.

The human body consists mostly of water, accounting for 50-70% of total body weight. Water is also the major component in body fluids, like blood and cerebrospinal fluid, and binds to large molecules like proteins.

The water molecule itself is composed of two hydrogen atoms and one oxygen atom bound tightly together (**Figure 6**).

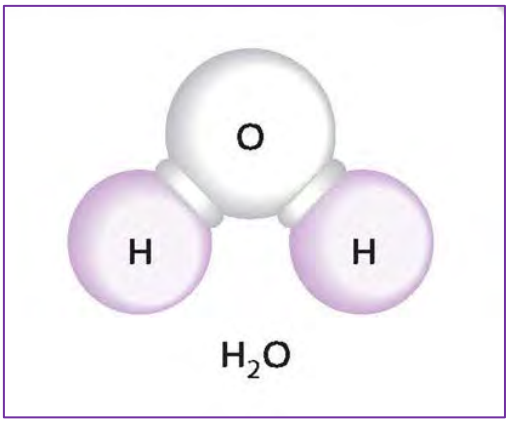

Figure 6. The molecular structure of water.

Hydrogen atoms are a primary element of body tissues, including fat, which is easily seen on MRI. There are ways to distinguish tissues containing water and fat using special MRI techniques, discussed later.

Other elements used to create MR images are sodium, phosphorus, and carbon. However, none are nearly as abundant in the body as hydrogen and do not generate the detailed high-quality images possible with hydrogen-based MRI. Consequently, these other elements are used rarely and almost exclusively for research purposes.

Hydrogen Structure

In addition to being one of the most plentiful elements in the body, the hydrogen atom is also the simplest. Hydrogen has an atomic number of 1, meaning that it contains a single proton and a single electron. The most common form of hydrogen does not contain neutrons. Because the nucleus consists of a single proton, the hydrogen atom is often just called a "proton" in the field of MRI, and standard MR imaging is frequently referred to as "proton MRI."

Property of Spin and Magnetic Characteristics

Along with its positive charge, the single proton in the hydrogen nucleus also has a physical property called **spin**. Because of the extraordinarily small size of a proton, the property of spin is not quite the same as what we would commonly think of as a spinning motion, but for this purpose it is useful and sufficiently accurate to visualize the behavior of hydrogen proton as a spinning top or gyroscope (**Figure 7**).

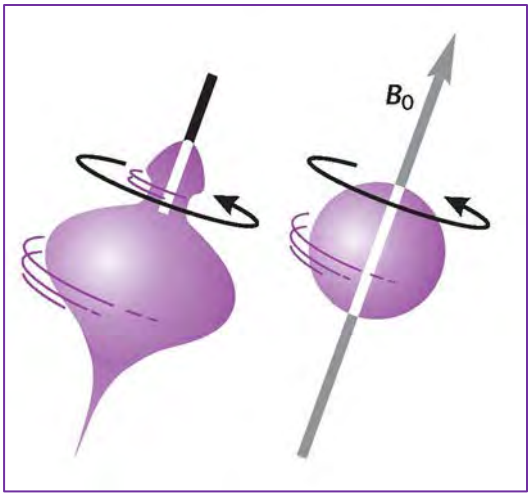

Figure 7. Protons may be visualized as spinning, charged particles. B_0 is the magnetic field vector that represents the direction and strength of magnetic force.

In this discussion, we will use the classical physics description of the behavior of the proton in the hydrogen nucleus as opposed to the quantum description, which is more accurate but also more complicated. The classical description is sufficient for describing how MR images are formed. Theoretically, any atom with an odd number of protons and/or neutrons also possesses the property of spin. However, since the hydrogen nucleus is the only one routinely used for clinical MRI, we will focus our discussion on the hydrogen proton.

Recall that the charged and spinning proton in the nucleus creates a small magnetic field or magnetic moment around the proton. In our simplified model, think of the spinning proton acting like a microscopic bar magnet: it will have both associated magnet strength (very small) and north and south poles (**Figure 8**).

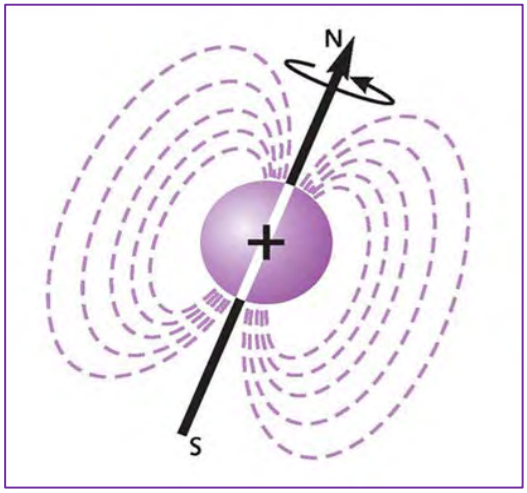

Figure 8. As a moving, charged particle, each proton is associated with a tiny magnetic field or magnetic moment.

The principles of physics assert that whenever there is an electrical charge moving or changing in some way, a magnetic force is generated.

When discussing its magnetic properties, the proton is frequently referred to as a magnetic dipole. The magnetic properties, arising from the spin property of the hydrogen nucleus, are what make hydrogen so useful for MRI. While other atoms possess these same magnetic properties, their quantity in the body and the quality of the images generated are very low as compared to hydrogen. Thus, the natural abundance of hydrogen in water and other body tissues makes it the dominant element of interest for MRI.

We now have a model of the hydrogen nucleus as an electrically charged and spinning particle. The principles of physics assert that whenever there is an electrical charge moving or changing in some way, a magnetic force is generated. This is one of the fundamental ways that the electrical and magnetic properties of matter are related.

POINTS for PRACTICE

1. What particles make up an atom? How does an atom become ionized?
The atom is the smallest unit of a chemical element and is made of up protons, electrons, and neutrons. The dense core of the atom, the nucleus, contains positively charged protons, as well as neutrons that carry no electrical charge. The number of protons determines the atomic number. Electrons orbit around the atom's nucleus and carry a negative electrical charge. The atom is said to be ionized when electrons are gained or lost in their orbits, leaving the atom with a net positive or net negative charge.

2. An electrically neutral atom has eleven particles in its nucleus and five particles orbiting the nucleus. How many protons does it have? neutrons? electrons?
5 protons, 6 neutrons, and 5 electrons

3. What chemical element is most often used for MRI imaging and why? What other elements can be used in MRI? The human body consists of 50-70% water, and hydrogen is the most abundant molecule in water. It is also the simplest, with an atomic number 1 (one proton and one electron and often no neutrons). Other elements used in MRI are sodium, phosphorus, and carbon.

4. Why can a positively charged hydrogen atom be described as a spinning top?
Atoms, particularly the hydrogen atom with its one proton, act like microscopic bar magnets, ie, a magnetic dipole with north and south poles.

The associated magnetic field is described by both a strength (magnitude) and a direction. Along with its positive charge, the single proton in the hydrogen nucleus also has a physical property called spin. Because of the extraordinarily small size of a proton, the property of spin is not quite the same as what we would commonly think of as a spinning motion, but for this purpose it is useful and sufficiently accurate to visualize the behavior of hydrogen proton as a spinning top or gyroscope.

POINTS for PRACTICE

1. Describe the magnetic field B_0.

2. What is bulk net magnetization? What symbol is used to describe it?

3. When a patient lies on the MRI table, the hydrogen nuclei in the body tissues respond to the magnetic field of the MRI scanner. What is the alignment of hydrogen nuclei at equilibrium? Why is this important?

4. What is precession? Describe the path that a precessing proton takes.

5. What is the Larmor frequency?

ATOMS AND MAGNETIC FIELDS

We have learned about the basic structure and properties of atoms, noting that some atoms, particularly the hydrogen atom with its one proton, act like microscopic bar magnets or magnetic dipoles. In this section, we describe how the magnetic properties of the proton interact with other magnetic fields produced by the MRI scanner to generate a signal that ultimately can be used to form an image.

Interactions between Atoms and Magnetic Fields

Recall that a magnetic dipole has with it an associated magnetic field. The magnetic field is described by both a direction (magnitude) and a strength. This holds true for the magnetic dipole generated by a single hydrogen proton. In body tissues, there are trillions and trillions of hydrogen protons acting as magnetic dipoles. The effects and measurements acquired in MRI are a result of the effect of all of the hydrogen protons combined.

Under ordinary conditions, the magnetic dipoles of hydrogen nuclei are oriented in random directions (**Figure 9**). Typically for every magnetic dipole pointing in a given direction in a sample, there is another pointing in the opposite direction. Together these magnetic fields cancel one another out, resulting in no net magnetization.

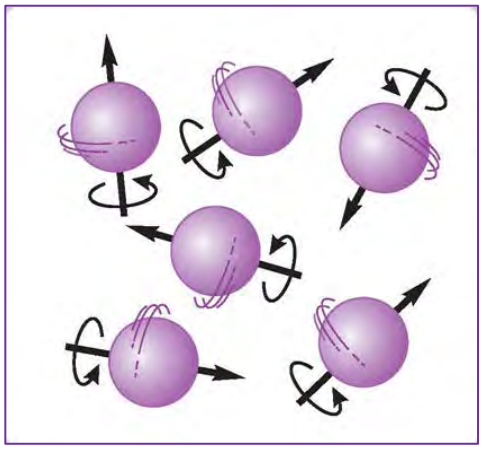

Figure 9. In the absence of a magnetic field, the magnetic dipoles of the hydrogen nuclei (protons) are randomly aligned.

Magnetic Field in MRI

The symbol $\boldsymbol{B_0}$ ("B-zero" or "B-naught") describes a magnetic field vector that represents the strength and direction of magnetic force. The zero subscript is used to distinguish this magnetic field from the other applied magnetic fields used in magnetic resonance imaging.

In the absence of a strong external magnetic field, the net magnetization of the hydrogen nuclei in body tissues typically adds up to zero. This behavior changes when the hydrogen protons are placed inside a strong, external magnetic field, much stronger than that generated by either the earth or a typical bar magnet.

Comparison of an MRI Magnet to Other Common Magnetic Fields

The main component of an MRI scanner is a very high-strength magnet. The usual unit of measure of a magnetic field is called **tesla**, named in 1960 for Nikola Tesla to honor his contributions to the field of physics. The earth's magnetic field is approximately 50 microtesla (50 millionths of a tesla), enough to deflect the needle of a compass but little else. A common refrigerator magnet can be up to 100 times stronger or 5 millitesla. By comparison, a typical MRI scanner today creates a magnetic field strength of 1.5 tesla, also written as 1.5T. This means the B_0 field at 1.5T would be 300 times stronger than a refrigerator magnet and 30,000 times stronger than the earth's magnetic field (**Figure 10**).

Commonly available MRI scanners in clinical settings use magnetic field strengths ranging from 0.2 to 3.0T, with even stronger B_0 fields used in research labs.

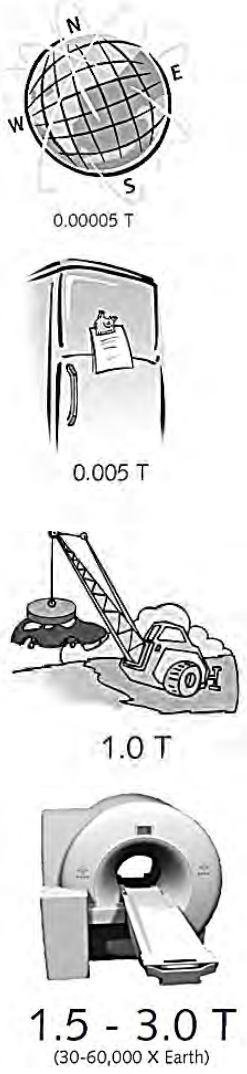

Figure 10. Examples of magnetic force.

Types of MRI Magnets

A **permanent magnet** for MRI is essentially like a huge refrigerator magnet. It is made of ferromagnetic metal, having its own permanent magnetic field. A permanent magnet is always "on" and requires no additional power or cooling to create the magnetic field. Permanent magnets for MRI are extremely heavy and therefore cannot be used for magnetic fields beyond about 0.6T. There also tend to be variations in the magnetic field over the bore of the permanent magnet that can lead to problems with image quality.

Resistive magnets are large electromagnets that create a magnetic field by applying an electrical current to a large coil of wire wound around an air or iron core. The magnetic field is generated and "on" only as long as this current is being applied and therefore can be turned off. The amount of electricity required is very large — on the order of 50 kilowatts — and therefore requires considerable cooling, making such magnets expensive to operate. These factors limit resistive magnets to field strengths below 0.7T. Permanent and resistive magnets are typically used only in open MRI systems and a few specialty systems such as extremity scanners.

The majority of MRI scanners installed in the United States utilize a **superconducting magnet** to generate the main magnetic field. Like a resistive magnet, the field is generated by the flow of electrical current in a coil of wire. However, the wire is made of special superconducting materials that have no electrical resistance when cooled to extremely low temperatures by liquid helium. Once the current is applied, no additional energy is required to maintain the magnetic field as long as the temperature is sufficiently low. Superconducting magnets can be made at much higher field strengths than resistive or permanent magnets. Fields of 1.5T are the most common in clinical practice, although 3.0T magnets are becoming more popular in new installations. Magnets up to 10.0T are used in human research.

Effect of a Large External Magnetic Field on Hydrogen Protons

Alignment

When a hydrogen nucleus is positioned in a strong magnetic field (an MRI scanner), it tends to line up in one of two states: spin-up or spin-down, also referred to as **parallel** or antiparallel alignment. These two states correspond to low-energy or high-energy states, respectively. If we represent the external magnetic field of the MRI scanner as a vector, B_0, a hydrogen nucleus and its magnetization will either line up pointing in the same direction as B_0 (the parallel orientation, corresponding to the lower-energy state), or it will line up pointing in the opposite direction of B_0 (the antiparallel orientation, corresponding to the higher-energy state).

When the tremendous numbers of hydrogen nuclei that make up any quantity of tissue are considered together, they will interact in complex ways. The net effect can be simply described such that the number of nuclei that line up in the parallel orientation is slightly greater than the number of nuclei in the antiparallel orientation (**Figure 11**). When added, the total magnetization of all of the antiparallel nuclei is canceled out by an equal number of parallel nuclei.

This leaves a small number of parallel (lower energy) nuclei that are not canceled out, resulting in a small overall magnetization pointing in the direction of B_0. In this alignment, the hydrogen nuclei are at **equilibrium** or in a resting state. In the absence of any other application of energy, the hydrogen nuclei of the tissues remain in an equilibrium state indefinitely.

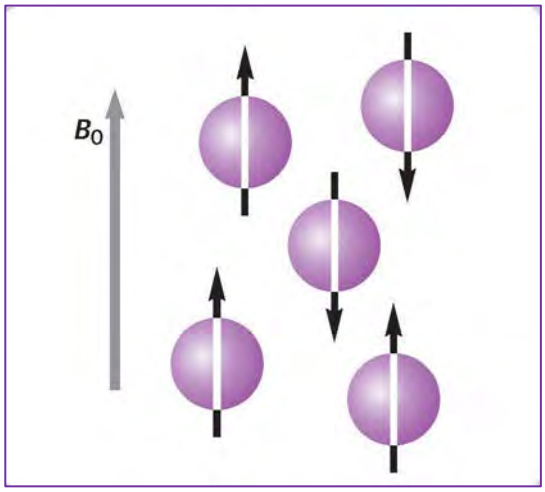

Figure 11. When placed in a strong magnetic field, slightly more than half of hydrogen nuclei align parallel to the field.

Bulk Net Magnetization

When a patient lies on the MRI table, the actual number of extra nuclei in the parallel orientation is only a few protons per million, but this is enough to produce a magnetization that can be used to generate an image using MRI techniques. It is important to note that the fraction of parallel nuclei that generates this magnetization increases approximately proportionately as the external B_0 strength increases. The sum of the magnetizations of these extra parallel nuclei, or magnetic moments, is known as the **bulk net magnetization**, symbolized by the vector *M* (**Figure 12**). We will continue to use the quantity *M* to describe changes in this bulk net magnetization as we explain each step of the magnetic resonance imaging path.

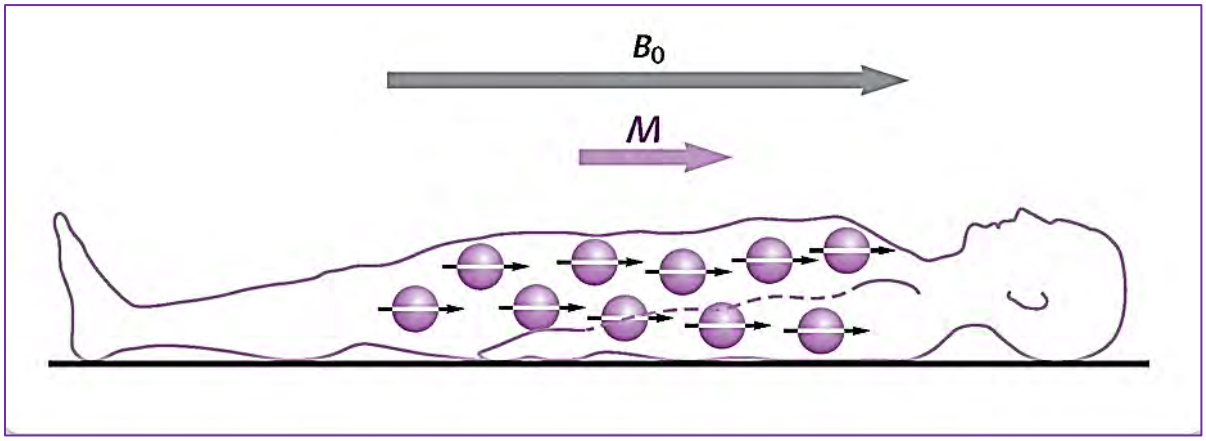

Figure 12. The bulk net magnetization vector *M* arising from a sample of protons in an external magnetic field (B_0) points in the same direction as the external field.

Dynamic Behavior of Hydrogen in a Large External Magnetic Field

Precession

We described how a bulk net magnetization is created in tissues by placing them in a large magnetic field so that the magnetic dipoles align in a known direction. This is necessary for generating a signal with MRI, but it is not yet sufficient for imaging. To create a signal that can be measured, recorded, processed, and displayed as an image, we must also introduce another property called **precession** of the hydrogen nuclei. Recall that the motion of the hydrogen nuclei is like a top or gyroscope, spinning around its own central axis. Continuing with this analogy, if we can force the nuclei to be tipped away from their "upright" direction aligned with the B_0 field, a "wobbling" motion begins **(Figure 13**). In addition to continuing to spin around its own axis, the top will more slowly rotate around the upright position, no longer perfectly aligned. This motion — precession — traces out a cone-shaped path around its original direction of alignment (**Figure 14**).

Consider a large group of nuclei in the parallel orientation; the nuclei will follow the same path but may be out-of-sync with each other such that each may be oriented at a different location around the cone. By adding together the vectors (*M*) that represent the individual magnetic moments of each nucleus, we arrive at the bulk net magnetization for the sample. In **Figure 15**, the total of these magnetic moments still adds up to a bulk net magnetization direction *M* along B_0 since the vectors are evenly distributed around the cone such that the parts of the vector pointing perpendicular to the B_0 direction cancel each other out.

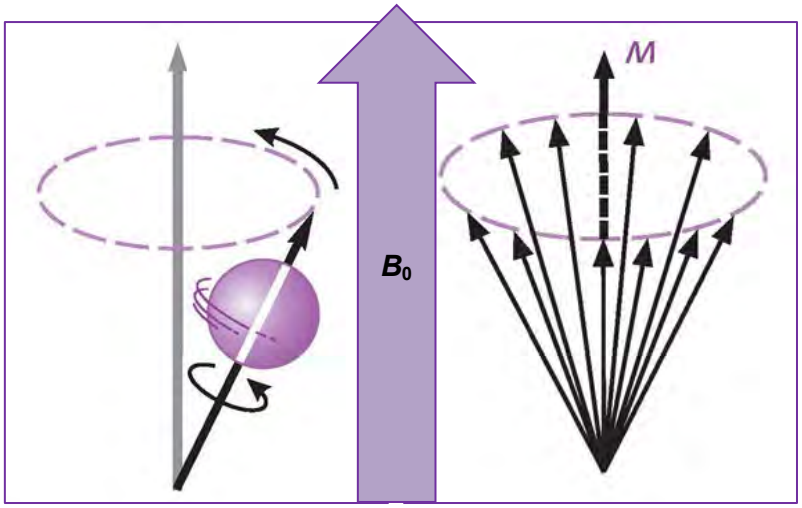

Figure 13. A proton's precession resembles the wobble of a top as it spins.

Figure 14. Protons precess in a cone-shaped path around an axis parallel to the direction of the magnetic field.

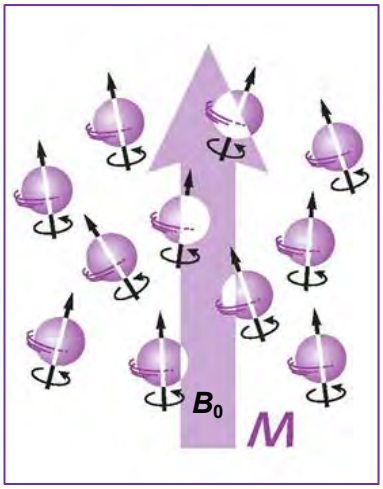

Figure 15. The bulk net magnetization vector *M* represents the sum of the magnetic moments of individual protons.

Rate of Precession Depends on Magnetic Field Strength

The rate of precession of the hydrogen protons is determined completely by the strength of the magnetic field in which the nuclei are placed. In MRI, since all of the nuclei are placed in the same large magnetic field, the rate of precession is the same for all of the nuclei. The rate is measured in terms of **frequency**, which is the number of revolutions (cycles) per second that are traversed. The relationship between the frequency of precession and the magnetic field is described by the **Larmor equation**. This equation states that

$$f = \gamma B_0$$

In this equation, *f* represents the frequency in cycles per second (hertz), B_0 is the main magnetic field strength, and γ (gamma) stands for the **gyromagnetic ratio** (sometimes called the magnetogyric ratio). The gyromagnetic ratio is a physical constant unique to each type of atom, meaning that the value of γ for hydrogen is different from, for example, that of sodium or phosphorus. The actual values of γ for different nuclei have been determined by experiments, since they cannot be determined from other properties.

As you see from the equation, the frequency of precession and the magnetic field are directly proportional, so as the magnetic field strength increases, so does the frequency. The frequency denoted by *f* depends on both the magnetic field strength and the type of nuclei being considered. For hydrogen, the value of γ is 42.575 MHz/T (megahertz or millions of hertz per tesla). For a typical 1.5T scanner, the Larmor frequency is therefore 42.575 MHz/T x 1.5T = 63.86 MHz. In comparison, for a 0.5 T magnet, the Larmor frequency for hydrogen is about 21.29 MHz, while at 3.0 T the Larmor frequency is 127.72 MHz, demonstrating that field strength and Larmor frequency are proportional. MR frequencies fall near those used by FM radio and are in the radiofrequency range (RF range) of the electromagnetic energy spectrum.

Perturbing the Protons

In order to induce precession, the protons must be **perturbed** or “pushed out” of their resting or equilibrium state. Otherwise, the protons remain in the lower energy equilibrium and will not provide any signal that can be used to create an image. To induce this change requires energy transmission into the body in the form of electromagnetic energy in the RF range to make the protons precess, a process called **excitation**.

We have now discussed not only the magnetic characteristics of a single hydrogen nucleus but the behavior of these nuclei in bulk in the presence of a large external magnetic field. The B_0 field aligns the nuclei to produce a bulk net magnetization in the direction of the B_0 field. Additionally, if the nuclei are tipped away (perturbed) from this alignment, they will precess around the direction of the B_0 field at a characteristic frequency known as the Larmor frequency, which is determined by both the type of atom and the strength of the magnetic field.

POINTS for PRACTICE

1. Describe the magnetic field B_0.
B_0 is a magnetic field vector that represents the direction and strength of magnetic force. The zero subscript is used to distinguish this magnetic field from any other applied field, for example, the B_1 field. B_0 represents the main magnetic field used in MRI, usually in the range of 0.2 to 3.0 tesla for clinical scanners.

2. What is bulk net magnetization? What symbol is used to describe it?
Bulk net magnetization is the sum of the magnetizations of excess parallel nuclei and is symbolized by the vector *M*. At typical MRI magnet strengths, the actual number of excess nuclei in the parallel orientation is only a few protons per million but is enough to generate a magnetization that can be used to generate an image. The proportion of parallel nuclei that generates this magnetization increases as the external magnetic field, B_0, increases.

3. When a patient lies on the MRI table, the hydrogen nuclei in the body tissues respond to the magnetic field of the MRI scanner. What is the alignment of hydrogen nuclei at equilibrium? Why is this important?
At equilibrium, a slight majority of hydrogen nuclei align their magnetic dipoles parallel to the magnetic field, which creates a net magnetization. These changes in the alignment of magnetic dipoles are necessary to generate the signal detected in MRI.

4. What is precession? Describe the path that a precessing proton takes.
Precession is a wobbling type of rotation performed by a magnetic dipole that is not aligned exactly parallel (or antiparallel) to an external magnetic field. The path of the precessing proton can be described as cone-shaped around its original direction of alignment.

5. What is the Larmor frequency?
This is the frequency at which the nuclei precess within the main magnetic field and is proportional to the magnetic field; the Larmor equation is $f = \gamma B_0$.

POINTS for PRACTICE

1. Where does MR imaging fall on the electromagnetic spectrum?

2. Illustrate the relationship between frequency and wavelength.

3. How is the equilibrium state of the proton perturbed?

4. Describe the role of the flip angle. What are the two primary types of flip angle used in MRI?

EXCITATION

Electromagnetic Waves

Electromagnetic energy results from a combination of electric and magnetic fields that travels together through space at the speed of light, approximately 186,000 miles per second (300,000,000 meters per second). While all types of electromagnetic energy share this description, there are many different categories of electromagnetic energy. The full range of these categories forms the **electromagnetic spectrum** (**Figure 16**).

Examples of electromagnetic energy include x-rays, visible light, and radio waves. These types of electromagnetic energy differ in the range of frequencies or wavelengths by which they are characterized and the amount of energy they carry. When electric and magnetic fields are combined, they continuously oscillate and can be described as an electromagnetic wave. Most often they are shown as sine waves (**Figure 17**). A sine wave (pronounced “sign”) is an s-shaped, oscillating wave with a repeating pattern.

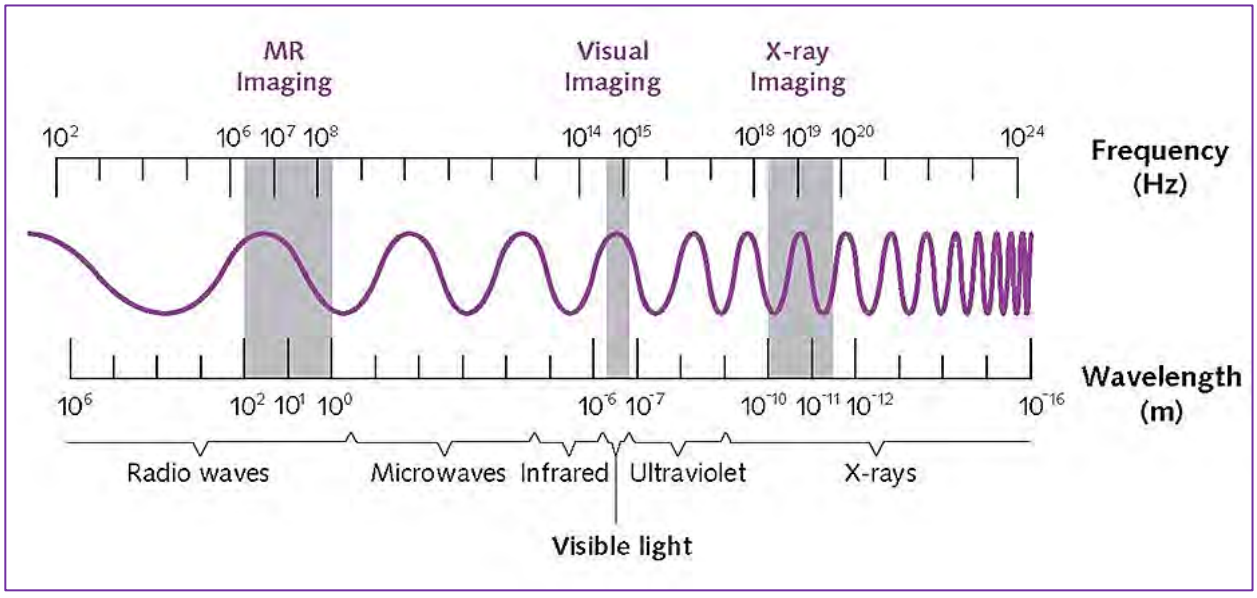

Figure 16. The electromagnetic spectrum consists of different types of electromagnetic energy.

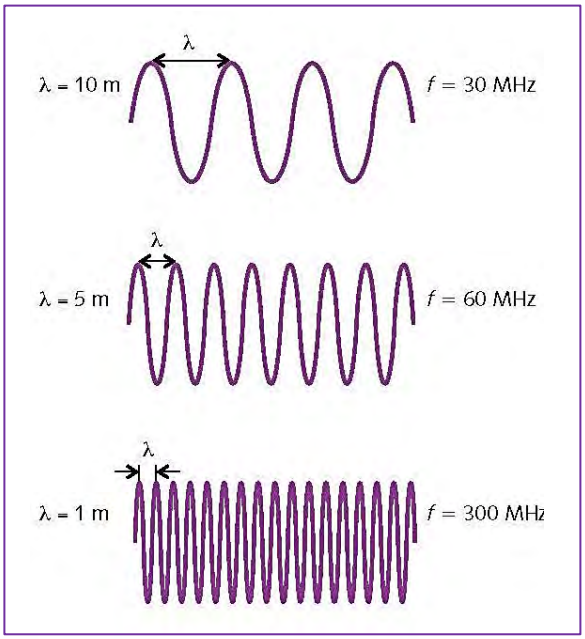

Figure 17. Three sine waves with different wavelengths (λ) and frequencies (f).

Frequency and Wavelength

Two important characteristics of a wave are **frequency** and **wavelength**. Frequency is defined as the number of times the wave pattern is repeated or the number of cycles the wave goes through in a given amount of time; it can also be described as the number of wavelengths passing by a given point in a given amount of time. Most often the unit of time is in seconds, and the frequency is expressed as the number of cycles per second or **hertz** (Hz).

Wavelength is defined as the distance between the two nearest corresponding points on the wave. Measuring corresponding points between the peaks, the valleys, or any other point yields the same result. Because the speed at which the wave travels is constant (the speed of light), the following equation relates frequency and wavelength:

$$f=c/\lambda$$

Here, *f* represents the frequency measured in hertz, *c* is the speed of light measured in meters per second, and *λ* (lambda) stands for the wavelength measured in meters. High frequency electromagnetic waves have a short wavelength and low frequency waves have a long wavelength, but they all travel at the same speed.

Electromagnetic Spectrum

There are many types of electromagnetic waves, and they are distinguished by their range of frequencies or the range of their wavelengths. The total range of frequencies and wavelengths form a continuous spectrum (refer to Figure 16). Radio waves have a much lower frequency (longer wavelength) than that of visible light waves; visible light waves have lower frequencies (longer wavelengths) than those of x-rays.

The frequency of an electromagnetic wave also relates to the quantity of energy it carries. The higher the frequency of an electromagnetic wave, the more energy it contains. Equivalently, the shorter the wavelength of an electromagnetic wave, the more energy it carries.

Radio waves are low-energy electromagnetic waves and fall in the same range of those used for radio and television broadcasts and other communications systems. MRI uses these radio waves to create images, which have a much lower energy than x-rays.

When energy in the form of electromagnetic waves is directed at a substance at a particular frequency, the energy is absorbed by the tissue and then emitted back with a signal that can be detected and recorded. This is the basic principle used to create MR images. Because the energy used in MR imaging is in a lower range of the electromagnetic spectrum, there is not the risk of tissue changes or damage as there is with x-ray-based imaging modalities.

General Principles of Excitation

We have learned that the large B_0 field of the MRI scanner causes the resting or equilibrium state of the hydrogen nuclei to be such that there are a few more protons oriented with the B_0 field than protons aligned against the field in the opposite direction. This produces a bulk net magnetization oriented in the same direction as the B_0 field. The direction of the B_0 field is also described as the **longitudinal** axis as it usually points the "long way" down the tube of a closed-bore magnet.

Energy In, Energy Out

We can change or perturb the nuclei from equilibrium by transmitting energy into tissue containing hydrogen nuclei, causing the nuclei to jump to a higher energy state. However, the correct type of energy is required to effect a change that results in a measurable signal.

> The release of the absorbed energy is the source of the signal in MRI.

The type of energy required to perturb the equilibrium of the nuclei comes from a short burst of electromagnetic energy in the form of a radiofrequency or **RF pulse**. The RF pulse must be transmitted at just the right frequency, that is, at the **Larmor frequency**, the frequency at which the hydrogen nuclei precess when disturbed from their alignment in the magnetic field. Recall that the Larmor equation is expressed by $f = \gamma B_0$. Energy at this frequency is absorbed by some of the nuclei, causing them to jump to the higher energy state and begin precessing, changing the bulk net magnetization. Once the RF pulse is turned off, these nuclei release the absorbed energy as they return to their lower energy equilibrium state. The release of the absorbed energy is the source of the signal in MRI.

Resonance

The Larmor frequency is also known as the resonant frequency of the hydrogen nuclei because energy absorbed by the nuclei at this frequency causes them to **resonate**.

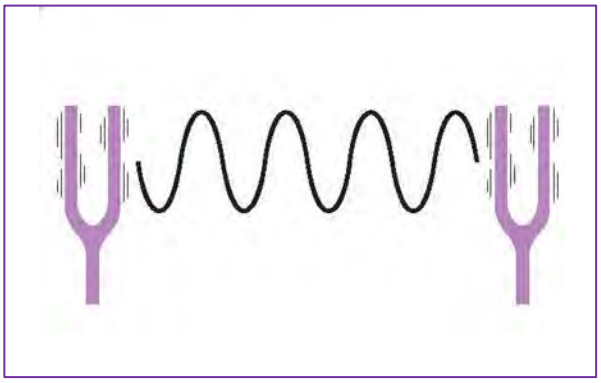

Figure 18. The second tuning fork vibrates after the first is struck because both are tuned at the same resonant frequency.

Resonance is defined as an object or material vibrating at a self-reinforcing "natural" or preferred frequency. These vibrations can be persistent and long-lasting. They also grow rapidly in strength as more energy is absorbed. Think of a tone being played at the resonant frequency of glass. If played long enough and loud enough, the vibration of the tone can cause the glass to break.

Another way to describe resonance and the transfer of energy involves a pair of tuning forks. Take two identical tuning forks and strike one to start it vibrating. Note that the other tuning fork begins to vibrate, too. The energy in the first tuning fork is transmitted as sound to the second tuning fork where it is absorbed and starts the second tuning fork vibrating (**Figure 18**). If the tuning forks are not tuned to the same note, the vibration of one tuning fork will have no effect on the other. The principle of resonance in MRI can be described by this same concept. Transmitting energy at just the right resonant frequency will cause the nuclei to absorb that energy.

Excitation: RF Pulses

Hydrogen nuclei exhibit a resonance condition at the Larmor frequency, and energy transmitted into the nuclei at this frequency causes them to move away from their equilibrium state and to absorb energy.

In MRI, the required energy is transmitted in the form of an RF pulse. Recall that electromagnetic energy consists of a combination of oscillating electric and magnetic fields. By transmitting an RF pulse, an additional oscillating magnetic field is introduced into the system. The magnetic field associated with an RF pulse is sometimes called the $\boldsymbol{B_1}$ **field**. The orientation of this oscillating magnetic field is in a direction perpendicular to the main B_0 field, the longitudinal axis, but the strength of the B_1 field is much smaller, on the order of microtesla. The orientation of the B_1 field is also referred to as being in the **transverse** direction.

Effect of Excitation on Spins

While the RF pulse is activated, the overall magnetic field experienced by the hydrogen nuclei is now the total of the B_0 and B_1 fields and therefore points in a slightly different direction (**Figure 19**). As the nuclei tend to align along the direction of a magnetic field, this causes the nuclei to tilt or "tip" slightly away from the main B_0 field direction and begin precessing at the Larmor frequency, the same frequency at which the RF pulse oscillates, creating the resonance condition. The match of the Larmor and RF frequencies causes the nuclei to further rotate and tip away from the B_0 axis.

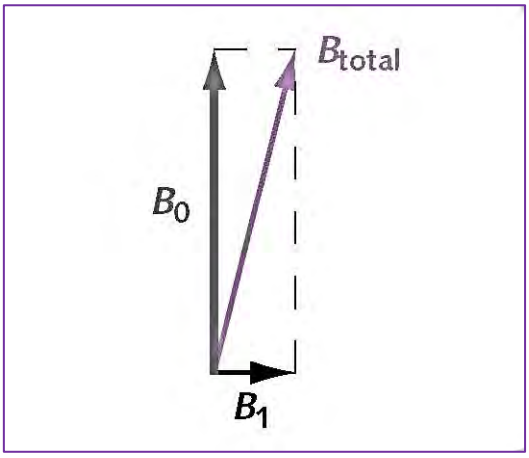

Figure 19. Total of B_0 and B_1 fields.

The rotation creates a spiral pattern as the direction of the magnetization simultaneously precesses (rotates around the B_0 axis) and is tipped farther away from the B_0 axis (**Figure 20**).

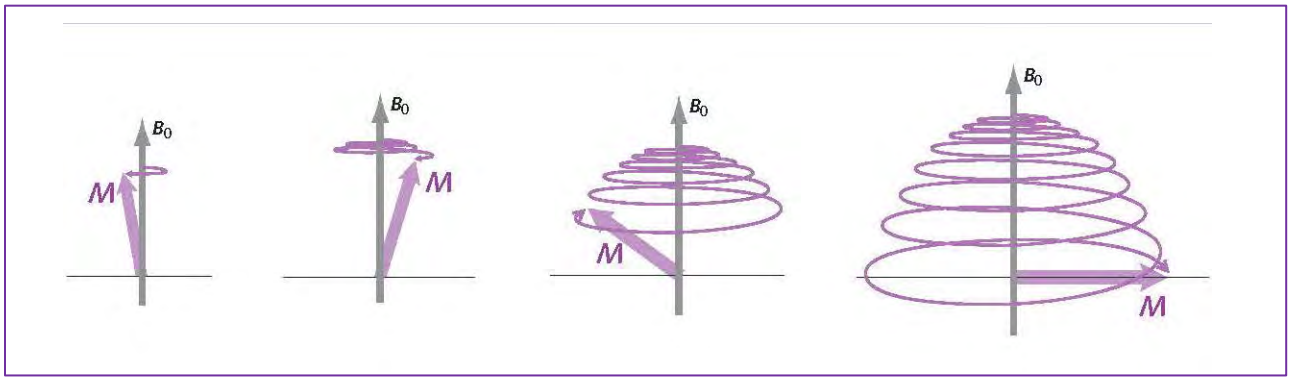

Figure 20. The match of the Larmor and RF frequencies causes the nuclei to rotate and tip away from the B_0 axis.

Flip Angle

The amount of the tip of the magnetization is measured in terms of the angle between the original B_0 axis and the angle of precession, called the **flip angle** (**Figure 21**). Equivalently, this is the angle between the angle of precession and the longitudinal axis. The overall effect is to change the orientation of the net magnetization from the longitudinal direction into the transverse plane. Immediately after excitation, the magnitude of the net magnetization is the same, but the direction that it points will have moved towards the transverse plane.

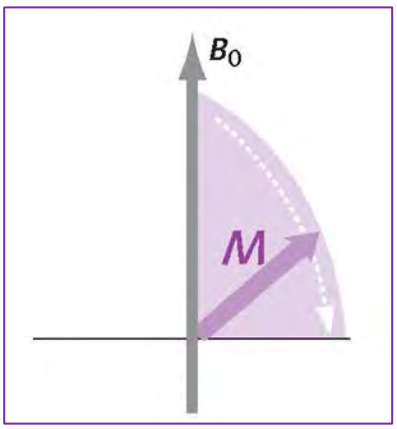

Figure 21. Flip angle. Magnetization M tipped away from the B_0 direction by an RF pulse.

Since the transmitted B_1 magnetic field is the same throughout the sample, the changes in the magnetization and the precession of all of the nuclei are synchronized. This means that all of the nuclei precess in unison, called **phase coherence** (**Figure 22**). When the RF pulse (B_1 field) is turned off, the nuclei continue to precess for a time but will gradually get "out of sync" and lose their phase coherence **(Figure 23**).

As just described, the RF pulse causes the nuclei to rotate away from the longitudinal orientation towards the transverse direction. This means that the bulk net magnetization in the longitudinal direction is also reduced below what it was in the equilibrium state.

The amount of change in the bulk net magnetization and in the flip angle away from the longitudinal B_0 direction is dependent on the strength and duration of the RF pulse. These dependencies are proportional, that is, if the duration of an RF pulse is doubled, the flip angle also doubles. Likewise, if the same duration RF pulse is transmitted with twice the strength, it doubles the flip angle.

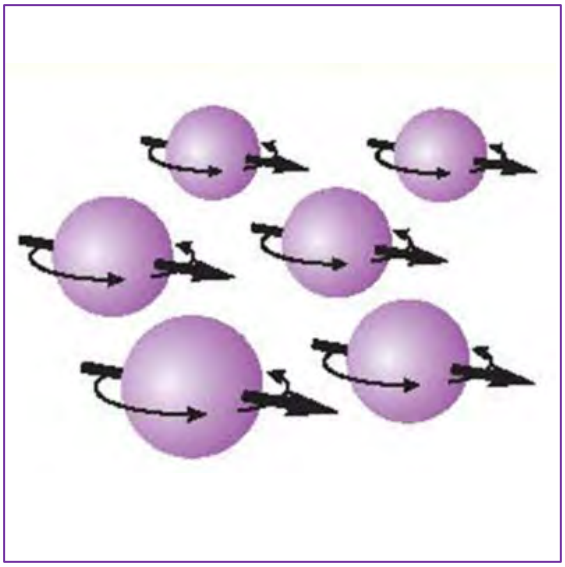

Figure 22. Phase coherence. Hydrogen nuclei precessing in unison.

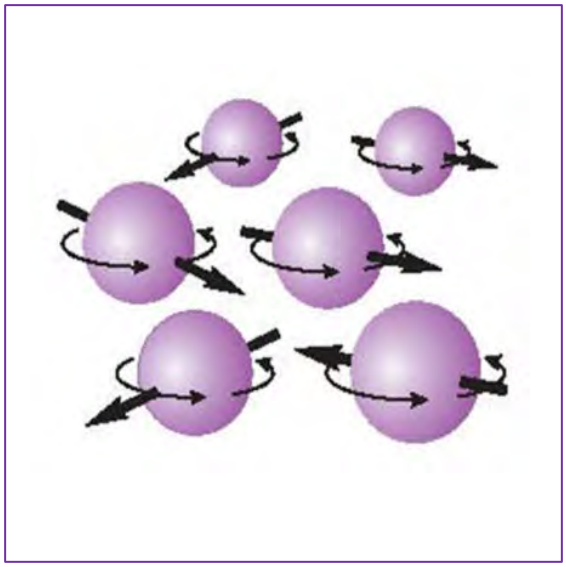

Figure 23. Out of phase. Hydrogen nuclei losing phase coherence.

90° and 180° RF pulses

Two categories of RF pulses are worthy of special mention because they are commonly used in MRI: the 90° and 180° RF pulses, named for the flip angle they each produce.

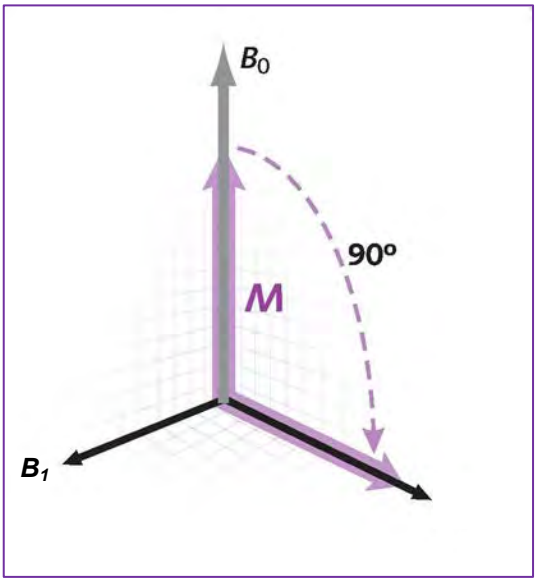

Figure 24. A 90° RF pulse rotates the bulk net magnetization by 90°.

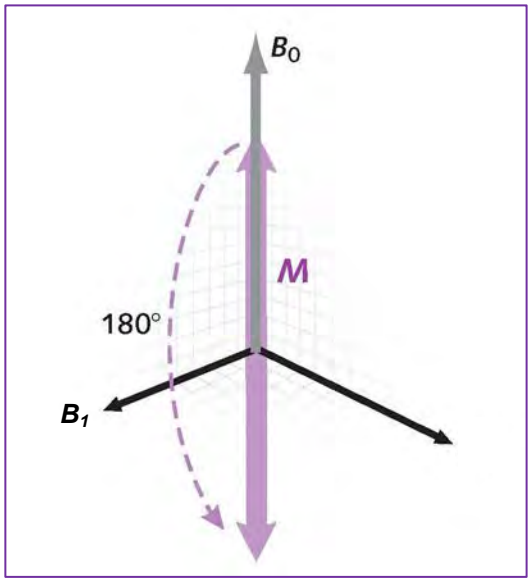

Figure 25. A 180° pulse rotates the bulk net magnetization by 180° so that it points in the opposite plane.

A 90° RF pulse causes the bulk net magnetization to completely rotate from the longitudinal axis (the orientation of the B_0 field) into the transverse plane (the direction in which the B_1 field is oriented); this is perpendicular to the static magnetic field (**Figure 24**). All of the magnetization along the longitudinal direction prior to the pulse is transferred to the transverse direction, leaving no magnetization along the longitudinal direction. As we will learn, the transverse component of the magnetization is what is measured as the signal in MRI, and the 90° RF pulse generates the maximum amount of signal.

The 180° RF pulse rotates the bulk net magnetization to point in the opposite direction (**Figure 25**). The magnetization along the positive longitudinal direction prior to the RF pulse will be rotated to point along the *negative* longitudinal axis. Here, none of the magnetization that starts along the longitudinal axis is moved into the transverse plane, so a 180° RF pulse by itself does not generate additional transverse magnetization or additional signal.

So why is the 180° RF pulse useful if it cannot generate any signal? First, it is used to restore some of the phase coherence in excited spins. Secondly, 180° pulses are applied to alter the appearance of tissues based on differences in how the magnetization returns to equilibrium.

POINTS for PRACTICE

1. Where does MR imaging fall on the electromagnetic spectrum?
Energy required for MRI falls at the low end of the electromagnetic spectrum. MRI uses radio waves to acquire an image, unlike x-ray or CT. Radio waves fall in the range of frequencies commonly used for communication broadcasting and are commonly referred to as radiofrequency, or RF, waves.

2. Illustrate the relationship between frequency and wavelength.
$f = c/\lambda$, where f is the frequency measured in hertz, c is the speed of light measured in meters/second, and λ is the wavelength measured in meters. High-frequency electromagnetic waves have a short wavelength, and low-frequency waves have a long wavelength. The frequency of MRI is much lower than that of x-ray or visible light.

3. How is the equilibrium state of the proton perturbed?
Perturbing the hydrogen proton from its equilibrium state requires a short burst of electromagnetic energy in the form of an RF pulse. The RF pulse must be transmitted at the right frequency, the Larmor frequency, which causes the protons to absorb energy and begin precessing.

4. Describe the role of the flip angle. What are the two primary types of flip angle used in MRI?
A flip angle describes the amount of tip of magnetization between the longitudinal axis, B_0, and the angle of precession. The two primary types of flip angle are 90° and 180°. A 90° flip angle changes the bulk net magnetization to completely rotate from the longitudinal to the transverse axis, that is, perpendicular to the static field. A 180° flip angle rotates the bulk net magnetization to point in the opposite direction, with no additional magnetization moved into the transverse plane.

NOTES

RELAXATION

POINTS for PRACTICE

1. Describe the process of relaxation and how it affects bulk net magnetization.

2. How are spin-lattice relaxation and spin-spin relaxation related?

3. Describe the time constants T1 and T2.

4. Dephasing and signal loss during transverse relaxation are affected by what two factors?

5. Differences in T1 and T2 permit the diagnostic use of MRI. Describe the reasons why.

Review of Excitation

Recall that an RF pulse causes the direction of the magnetization *M* to tilt away from the longitudinal direction and into the perpendicular transverse plane and that the flip angle describes the angle between the *M* direction and the longitudinal plane. If we describe the new magnetization *M* in terms of the precessing protons that are aligned in the longitudinal direction and the protons aligned in the transverse direction, the longitudinal magnetization is reduced, or the protons may even point in the opposite direction after the RF pulse.

The effects of the 90° and 180° pulses move the nuclei away from their resting condition due to the transmission of electromagnetic energy into the system. In the common 90° flip angle pulse, the amount of magnetization along the longitudinal direction goes to zero, with all the magnetization rotated into the transverse plane. For a 180° pulse, the new magnetization points in exactly the opposite direction. If we use compass directions as an analogy, magnetization that pointed “north” before the 180° pulse would point “south” after the pulse, and magnetization in the “east” direction would point “west” after the pulse.

When the RF pulse is turned off, the nuclei naturally begin returning to their resting state. Since the resting state is a lower energy state overall, the energy absorbed by the nuclei will in turn be emitted. This process is called **relaxation** and defined by both a loss of energy and loss of order or coherence in the system.

Return to Equilibrium — Relaxation

Two events occur simultaneously to return the nuclei to equilibrium where all of the magnetization *M* is along the longitudinal direction and all of the transverse magnetization vanishes. First, the magnetization in the transverse plane gradually decays to zero.

This process is called transverse relaxation and results from the loss of coherence set up by the excitation pulse. Secondly, longitudinal magnetization is restored to its resting state along the B_0 direction, known as longitudinal relaxation.

Longitudinal or Spin-Lattice Relaxation

One of the effects of excitation pulses is to cause some of the hydrogen nuclei to jump to a higher energy state as they absorb energy from the RF pulse. When the RF pulse is turned off, these nuclei lose the absorbed energy and return to their lower energy state, which is the preferred, more stable state. Because it is a lower energy state, the additional absorbed energy is released into the surrounding environment or lattice. This relaxation process is called **longitudinal relaxation** or **spin-lattice relaxation**.

T1 Time Constant

As more nuclei return to the lower energy state or relax, the number of low-energy nuclei increases compared to the high-energy nuclei until the original equilibrium state is achieved with the magnetization *M* along the longitudinal direction and the bulk net magnetization returning to its full equilibrium value or resting state.

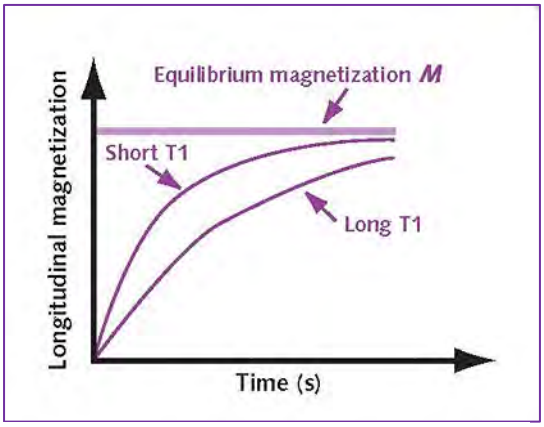

Figure 26. T1 recovery.

It is important to note that the nuclei do not all instantly return to the lower energy state when the RF pulse is turned off, nor do they all return at the same time. Individually, the nuclei randomly return from the higher energy state back to their lower energy state. However, because so many nuclei exist in any given tissue sample, the combined effect of their "behavior" on the bulk net magnetization is predictable because it represents the sum total or average of an extremely large number of nuclei.

The rate of restoration depends on the particular tissue being observed. This rate is described by the unit of time **T1**, a time constant that describes how rapidly the relaxation process occurs. T1 is usually reported in milliseconds and is typically in the range of 300 – 4000ms. A short T1 value means that the longitudinal magnetization is restored rapidly. A longer T1 value means that the magnetization recovers more slowly (**Figure 26**). Because different tissues have different T1 time constants, they can be made to appear differently on resulting images just by changing some of the scanner parameters.

FACTORS AFFECTING LONGITUDINAL RELAXATION

The primary factor determining the T1 time constant is the type of tissue. However, the strength of the B_0 magnetic field also has an influence. In general, tissues placed in a larger B_0 field will have a slightly longer T1 time, meaning it will take longer for the nuclei to return to their equilibrium state. This is important to note when comparing images acquired on different MRI scanners and when designing protocols for these scanners.

Transverse or Spin-Spin Relaxation

In order for the spins return to the equilibrium state, the magnetization in the transverse plane must vanish, just as the longitudinal magnetization must be restored. In theory, the loss of transverse magnetization must occur at least as fast as the longitudinal magnetization is regained, otherwise this would result in total magnetization increasing, which is not physically possible. In practice, for nearly all tissues the loss of transverse magnetization occurs much more quickly than the restoration of the longitudinal direction because of additional and more significant factors that cause it to fade away, which will be discussed later.

When the RF excitation pulse is turned on, the hydrogen nuclei spin coherently. The magnetization of all the nuclei adds up constructively, resulting in a bulk net magnetization in the transverse plane. In the case of a 90° excitation, the magnetization in the transverse plane will be equal to the longitudinal magnetization prior to the RF excitation. When the RF pulse is turned off, this coherence begins to be lost, and the sum total of the magnetization decreases.

The loss of coherence results from small random variations in the magnetic field in the hydrogen-containing molecules. Recall that the rate at which the nuclei precess is always directly proportional to the main magnetic field. If two nuclei experience slightly different magnetic fields, they spin at slightly different rates and begin to get out-of-sync with each other and become **dephased** (**Figure 27**). As a result, total magnetization decreases because the magnetization no longer adds up completely constructively.

One way to visualize this is to think of an auto race around a circular track. At the beginning of the race, all of the cars are lined up together and moving at the same rate behind the pace car so that the cars stay together and are coherent. Once the race begins, each car will drive at a slightly different speed. Their locations around the track begin to spread out, and the spacing gets wider over time, that is, the cars (nuclei) become dephased.

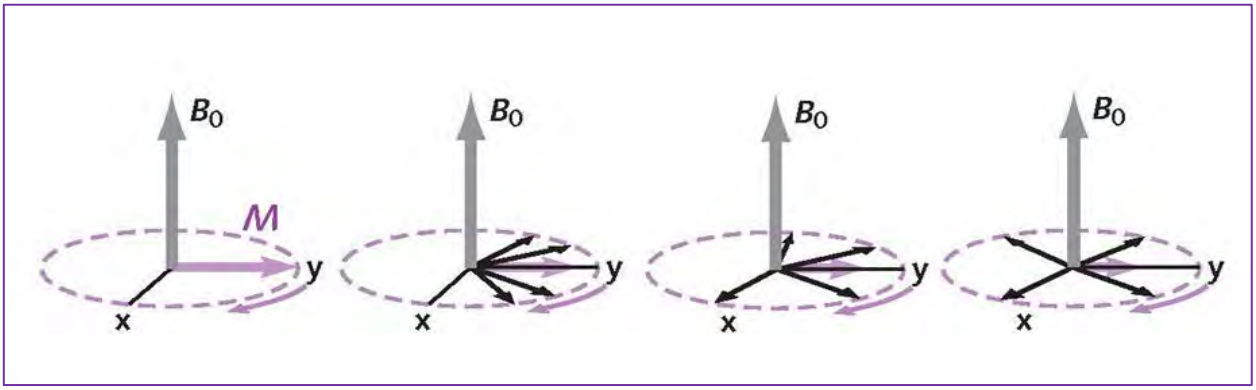

Figure 27. Loss of phase coherence or dephasing occurs as nuclei precess at different rates during transverse relaxation. Notice that the total magnetization *M* decreases.

As time goes on and the spin of the nuclei continue to dephase, the transverse magnetization continues to be reduced until it is eventually lost completely as the orientation of the spins becomes random. The process by which these microscopic field variations cause loss of coherence and transverse magnetization is called **transverse relaxation** or **spin-spin relaxation**. The measurement of transverse magnetization over time looks like the curve shown in **Figure 28**, decaying from its initial magnetization after the end of the RF pulse towards zero. The shape of this graph is particularly important because the signal detected and processed to create an image comes from transverse magnetization. Once transverse magnetization is lost, the signal to be recorded for imaging no longer exists.

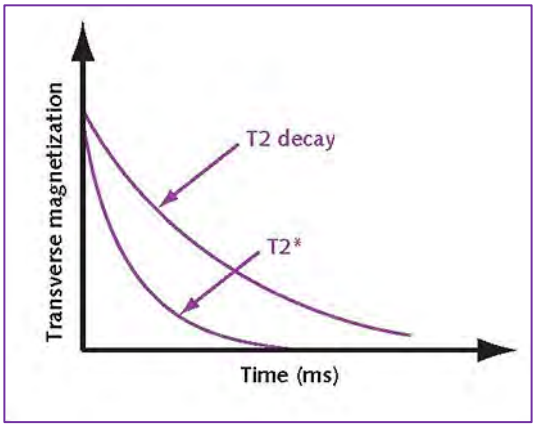

Figure 28. Transverse relaxation. T2* relaxation is always more rapid than T2 relaxation.

T2 Time Constant

Like longitudinal relaxation, the rate at which the signal is lost can be described by a time constant. For transverse or spin-spin relaxation, this time constant is called **T2** and is also measured in milliseconds, with a typical range of 30-150ms. A short T2 time means that the transverse magnetization is lost more quickly than it is for tissue with a longer T2 time. In practice, T2 is almost always shorter than T1 by a factor of around 10 or more.

Factors Affecting Transverse Relaxation — T2*

In a clinical scanner environment, two factors cause magnetic field variations to lead to dephasing and signal loss. The first is the chemical make-up of the tissue. The arrangement of atoms inherently causes small variations in the local magnetic field. These variations are specific to the molecule and account for some of the effects included in the T2 time constant.

The second factor comes from the interactions between the externally applied magnetic field and the tissue. Ideally, the B_0 created by the magnetic field would be perfectly homogeneous, that is, exactly the same everywhere. However, due to extremely small variations in the manufacturing of the magnet, as well as interactions with the type and arrangement of body tissues, for example, near the air pockets of the sinuses, the actual magnetic field experienced by the nuclei at a particular location could be slightly different than at other locations. These differences are referred to as **inhomogeneities**. Usually these variations are only a few parts per million (ppm) but are enough to cause dephasing and signal loss at a rate greater than from T2 relaxation alone. The combined effect of T2 relaxation and these additional factors is called **T2* relaxation (T two star)**, so named because the effect on the transverse magnetization is the same as purely T2 relaxation but occurs at a different and faster rate. Note the graph in Figure 28 illustrates both T2 and T2* relaxation effects.

Once transverse magnetization is lost, the signal to be recorded for imaging no longer exists.

T1 & T2 Relaxation and Imaging

T1 and T2 relaxation processes occur for all tissues that give off an MR signal. T1 and T2 times are different depending on the tissue type, that is, for different populations of hydrogen nuclei. However, T1 and T2 relaxation rates can be considered to be constant for a particular type of tissue.

For example, the hydrogen protons in pure water have a very long T1 relaxation time, while the hydrogen protons in fat have a much shorter T1 relaxation time. These characteristic variations in the response of different tissues to the magnetic fields permit us to discriminate one type of tissue from another on an MR image. It is also important to recognize that the main magnetic field strength affects specific relaxation times. In particular, the T1 relaxation time of most tissues is longer in a 3.0 T magnetic field than for a 1.5T magnetic field, while T2 relaxation times are usually about the same or slightly shorter. The appearance and contrast of different tissues may

therefore be slightly different on images taken at different field strengths. This should be considered when creating imaging protocols and comparing images from different scanners.

The difference in tissue response is what gives MRI its diagnostic ability and power. The differences in T1 and T2 are reflected in the signal intensities seen in the images produced, and these differences permit the radiologist to identify and interpret the appearance of tissue and arrive at a diagnosis. In addition, the appearance of tissues can be controlled by careful selection of parameters on the scanners, permitting imaging protocols to be designed to best highlight these differences for specific anatomy and conditions.

The difference in tissue response is what gives MRI its diagnostic ability and power.

We have now learned about both the process and set of conditions that the bulk net magnetization of hydrogen nuclei experience in a single MRI scan acquisition. The nuclei are initially at equilibrium with all magnetization along the longitudinal direction of the B_0 field. When an RF pulse is applied, the magnetization rotates into the transverse plane to a degree dependent on the duration and strength of the RF pulse.

After the RF pulse is turned off, the nuclei immediately begin the T1 (longitudinal) and T2 (transverse) relaxation processes. The T2 relaxation process causes the transverse magnetization to fade away to zero, while T1 relaxation restores the magnetization back along the longitudinal direction and returns to equilibrium. Both T1 and T2 relaxation rates are dependent on tissue type, as well as on variations in the magnet itself, allowing MRI to differentiate not only types of tissues but to distinguish between normal and diseased tissue.

NOTES

POINTS for PRACTICE

1. Describe the process of relaxation and how it affects bulk net magnetization.
Relaxation is defined by a loss of energy and increased randomness of the system as it returns to equilibrium. Resonating hydrogen nuclei undergo relaxation by emitting energy, dephasing, and returning to a parallel orientation to the main magnetic field. The bulk net magnetization vector *M* reflects these changes as alterations in longitudinal and transverse magnetization.

2. How are spin-lattice relaxation and spin-spin relaxation related?
Spin-lattice relaxation is the restoration of the longitudinal component of *M*, while spin-spin relaxation is the decay of the transverse component of *M*. After the RF pulse is turned off, the longitudinal component increases in magnitude towards its equilibrium value, while the transverse component decays to zero.

3. Describe the time constants T1 and T2.
T1 and T2 are time constants that characterize the rate of magnetic relaxation and are dependent on the particular tissue being observed. T1 describes how rapidly the relaxation process occurs in the longitudinal axis after the RF pulse is turned off. A short T1 value means that the longitudinal magnetization is restored rapidly; a longer T1 value means that the magnetization recovers more slowly.

Conversely, T2 is a time constant that characterizes the decay of the MR signal after the RF pulse is turned off. A short T2 time means that the transverse magnetization is lost more quickly than it is for a tissue with a longer T2 time. T2 must be shorter than T1 and may be shorter by a factor of 10 or more. Once the transverse magnetization is lost, the signal to be recorded for imaging no longer exists.

4. Dephasing and signal loss during transverse relaxation are affected by what two factors?

There are two factors that cause magnetic field variations leading to dephasing and signal loss. The first is the chemical make-up of the tissue. The arrangement of atoms inherently causes small variations in the local magnetic field. These variations are specific to the molecule and account for the effects included in the T2 time constant.

The second factor comes from the interactions between the externally applied magnetic field and the tissues. Due to differences in the manufacturing of the magnet and tissue arrangements (near air pockets for example), the actual magnetic field experienced by the nuclei at a particular location could be slightly different. Usually these variations are only a few parts per million, but this is enough to cause dephasing and signal loss at a rate > T2 alone. The combined effect of T2 relaxation and these two additional factors is called T2* relaxation. Its effect on transverse magnetization is the same as purely T2 relaxation but occurs at a different and faster rate.

5. Differences in T1 and T2 permit the diagnostic use of MRI. Describe the reasons why.
In physical terms, T1 and T2 are time constants that characterize the rate of magnetic relaxation. But for clinical purposes, these quantities are tissue characteristics that vary for different types of body tissue. The variance of T1 and T2 provides the MRI operator with a basis for generating images that reflect subtle differences between soft tissues, resulting in the demonstration of fine anatomical details. Additionally, images may be acquired under various conditions that highlight or minimize the influence of T1 and T2, adding to the power and flexibility of MRI.

FREE INDUCTION DECAY

POINTS for PRACTICE

1. Describe Faraday's law.

2. Name some variables that determine the amplitude of the free induction decay signal.

After Excitation

Recall that immediately after an excitation pulse, the magnetization of the nuclei is tipped into the transverse plane, which is perpendicular to the direction of the main magnetic field B_0. The magnetization precesses around the B_0 direction, spinning at the Larmor frequency. A visual analogy would be the spinner in a board game.

In MRI, the direction of the net magnetization changes extremely rapidly. In a typical 1.5T clinical scanner, this frequency is approximately 64 MHz. In other words, the magnetization rotates 64,000,000 times per second, or once every 15 nanoseconds. At 3T, this would be twice as fast. This rate of spinning falls within the RF range.

Faraday's Law

Faraday's law is a principle of physics that states that a spinning magnetization or any magnetization that changes over time can generate an electrical voltage in a nearby coil of wire; the voltage created in that wire coil can then be used to produce a measurable signal after excitation (**Figure 29**). A coil of wire acting like an antenna in a radio detects the rotation or precession of the magnetization and generates an electrical voltage. In essence, the energy absorbed during excitation is transmitted back into the coil. Note that Faraday's law requires a *changing* magnetization. When the magnetization is at equilibrium and pointing only in the longitudinal direction, there is no change and therefore no voltage created.

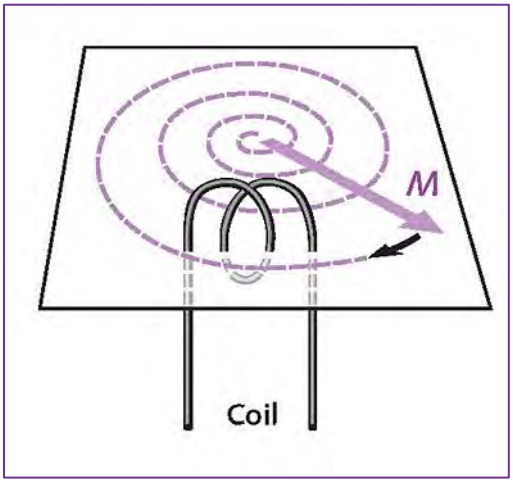

Figure 29. Precession of the net magnetization vector *M* within the transverse plane induces an electric voltage in an RF receiver coil.

Free Induction Decay

The recorded signal created by measuring the electrical voltage generated by the coil after an excitation is called the **free induction decay** (FID). This signal is quite weak, but when fed into the electronics of the scanner to be amplified and processed, it ultimately generates the signal used to form an image. The strength or **amplitude** of the FID generated by the coil depends on the strength of the magnetization *M* after excitation as well as other factors, such as the distance of the coil from the tissue and the frequency at which the magnetization precesses. A closer coil, such as a surface coil, gives stronger signal; a higher frequency, like that generated in a stronger main magnetic field, gives a stronger signal.

The term free induction decay describes the fact that the signal results from free precession of the magnetization while there is no external RF energy being applied.

The FID signal rapidly oscillates due to the precession of the magnetization. This oscillation occurs at the Larmor frequency within the RF range. **Figure 30** shows the signal that is recorded after such an excitation.

In addition to the oscillation of the signal, the signal strength gradually decreases over time due to T2 relaxation, causing the magnetization to decay.

It is now clear that the term free induction decay describes the fact that this signal results from free precession of the magnetization while there is no external RF energy being applied. This causes induction of voltage in the coil as the transverse magnetization decays due to T2 relaxation. The voltage continues to oscillate at the Larmor frequency even as the signal strength decreases.

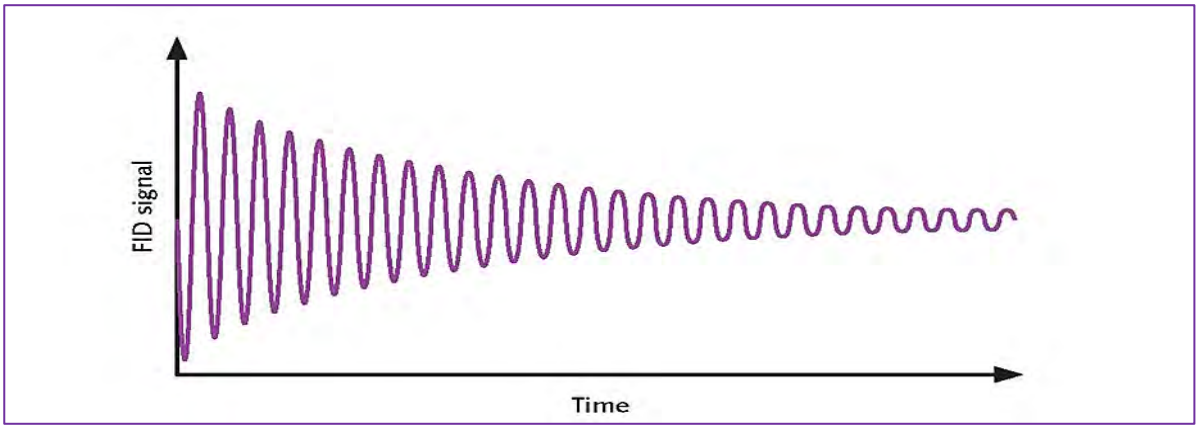

Figure 30. A graph of FID signal shows that its amplitude becomes smaller over time due to T2 relaxation as *M* returns to equilibrium.

The coils integrated into the MR system can either be the same coils used for excitation (a transmit/receive coil) or separate coils (a receive-only coil). These coils are produced in a variety of shapes and sizes, but their operating principle is the same. The FID is the basic signal created by excitation through the induction of voltage in coils, which is subsequently recorded by the MRI system. It provides information about the presence of hydrogen atoms but is not sufficient to create a useful image for clinical diagnosis. First, the FID does not carry any information about the location of the atoms used to form an image except that they are near the coil, and second, the signal generated does not provide sufficient information in the resulting images to differentiate between tissue types.

We have learned how changes in magnetization of a tissue caused by excitation can be detected and recorded. Rapidly changing magnetization creates an electrical voltage in a receiving coil that is measured and recorded. This recorded signal is the free induction decay, which is characterized by rapid oscillations due to precession and gradual decay due to T2 relaxation.

Further study is required to learn how a series of excitation pulses – pulse sequences – can be used to create images with particular characteristics that highlight the relaxation properties of different tissues. The arrangement and timing of these pulses are largely under the control of the scan operator, which gives MRI the flexibility to create many different types of images with the same scanner.

POINTS for PRACTICE

1. Describe Faraday's law.
Faraday's law is a principle of physics that states that a spinning magnetization (or any magnetization that changes over time) can create an electrical voltage in a nearby coil of wire. In MRI, this is the principle used to generate a measurable signal after excitation.

2. Name some variables that determine the amplitude of the free induction decay signal.

- distance from the coil
- frequency of precession
- strength of the bulk net transverse magnetization after excitation

SUMMARY

MRI has become one of the most widely available and routinely used examinations for assessing tissues and pathology throughout the body. Compared to CT and other imaging modalities, MRI provides unsurpassed versatility in its range of applications and clinical uses. Continuing innovations in MRI technology assure this will be true for decades to come. While many of the features on an MRI scanner are now automated, understanding and appreciating the basic principles of the physics of MRI is vital to your role as a radiologic technologist.

NOTES

GLOSSARY OF ABBREVIATIONS

B_0	the main magnetic field
B_1	magnetic field for RF transmission pulse oriented 90° to the main magnetic field
CNR	contrast-to-noise ratio
CNS	central nervous system
CSF	cerebrospinal fluid
DAQ	data acquisition
dB/dt	change in magnetic field per unit time
DWI	diffusion-weighted imaging
EPI	echo-planar imaging
f	frequency
FFE	fast field echo (same as gradient echo)
FID	free induction decay
FISP	fast imaging with steady-state precession (same as SSFP)
fMRI	functional MRI
FLAIR	fluid-attenuated inversion recovery
FOV	field of view
FS	fat suppressed (also fat sat)
FSE	fast spin echo
FT	Fourier transform
γ	gyromagnetic ratio
GBCA	gadolinium-based contrast agent
Gd	gadolinium
GRE	gradient echo (also gradient recalled echo)
Hz	hertz (cycles per second)
IR	inversion recovery
kHz	kilohertz
M	bulk magnetization vector
MHz	megahertz
MIP	maximum intensity projection
mm	millimeter

MRA	magnetic resonance angiography
MRS	magnetic resonance spectroscopy
ms	milliseconds
mT	millitesla
NMR	nuclear magnetic resonance
NSA	number of signal averages (also NEX)
NSF	nephrogenic systemic fibrosis
PC	phase contrast
PD	proton density
ppm	parts per million
RF	radiofrequency
SAR	specific absorption rate
SE	spin echo
SNR	signal-to-noise ratio (also S/N)
SSFP	steady-state free precession (same as FISP)
STIR	short-T1 or short-tau inversion recovery
T	tesla
T1	time for 63% of a tissue's longitudinal magnetization to recover
T1W	T1 weighting
T2	time for 63% of a tissue's transverse magnetization to decay
T2W	T2 weighting
T2*	time constant that characterizes the rate of transverse relaxation in an inhomogeneous magnetic field
TE	echo delay time; echo time
TI	inversion time; time to inversion
TOF	time-of-flight
TR	repetition time; time to recovery; recovery time
TSE	turbo spin echo
W/kg	watts per kilogram

GLOSSARY

aliasing
a common artifact caused when the field of view selected is smaller than the area of tissue being excited; also known as “wrap-around” or “wrap”

amplitude
the maximum magnitude or intensity of change in an oscillating variable

angiogram
an image of arteries and/or veins in the body. In MRI, angiograms are projection images created from multiple images acquired with flow-sensitive imaging protocols. Depending on the sequence selected, MRA can measure both flow of blood and its direction throughout the vasculature.

artifact
in the science of imaging, a substance or structure not naturally present in the imaged material but which appears in an image

averaging (AVG)
acquiring the same image multiple times, then adding the images together to improve quality. Also known as number of excitations (NEX) or number of signal averages (NSA).

B_0 (B-zero)
a magnetic field vector that represents the direction and strength of magnetic force, usually of the main magnetic field of the scanner; measured in tesla

B_1 (B-one)
the RF magnetic field applied at the resonance condition with frequency that is the same as the Larmor frequency

bandwidth
refers to a range of frequencies; this range helps determine slice thickness

bulk net magnetization (*M*)
sum of the individual magnetic moments of a group of magnetic dipoles

cardiac gating
synchronization of imaging with a phase of the cardiac cycle (image acquisition) between the patient’s heart beats in order to ‘freeze’ the heart motion

chemical shift
occurs when the chemical properties of a substance cause a shift in frequency at which it resonates as compared to other substances

coils
an electromagnetic device formed by winding one or more turns of wire or other conducting material around a form. In MRI, coils are used to generate magnetic fields (RF transmit coils and gradient coils, for example) and detect changing fields (receiver coils).

contrast-to-noise ratio (CNR)
the difference in intensity between two tissues of interest relative to the noise level

coordinate axes
set of perpendicular lines used as fixed references for determining the position of a point or a series of points; often designated as x, y, and z

crosstalk
the small amount of tissue outside the selected slice that may be excited by an RF pulse and therefore generate signal in the image or be saturated in an adjacent slice

dB/dt
the rate of change of the magnetic field per unit time. Because rapidly changing magnetic fields can induce electrical currents, this is an area of potential concern for safety limits.

dephasing
the “fanning out” of spins due to slight variations in the main or local magnetic field

diamagnetic
an element that is slightly repelled by a magnetic field, eg, helium, copper, and gold

diffusion-weighted imaging
an acquisition technique that generates images with intensities that depend in part on the microscopic motion of water molecules

dipole (magnetic)
a pair of north and south magnetic poles, separated by a finite distance

echo-planar imaging (EPI)
similar to fast spin or turbo spin echo, multiple lines of *k*-space are acquired after each excitation but at a much faster rate; the fastest of the gradient-echo-based scanning protocols, although resolution and image quality may be lower than that of standard images

echo time (TE)
time interval between the initial RF pulse and the echo of a pulse sequence; also echo delay time

echo train length
see turbo spin echo

electromagnetic spectrum
continuous series of different types of electromagnetic energy, ordered according to wavelength or frequency

equilibrium
state of rest or balance

Ernst angle
describes the flip angle that generates the largest amount of signal possible for a particular tissue T1 and pulse sequence TR combination. Named for the Swiss physical chemist Richard R. Ernst who was awarded the Nobel Prize in Chemistry in 1991.

excitation
to disturb the equilibrium of the precessing proton; to perturb

Faraday's law
the principle of physics that states that a spinning magnetization (or any magnetization that changes over time) can generate an electrical voltage in a nearby coil of wire. Named in honor of English chemist and physicist, Michael Faraday (1791-1867).

fast imaging with steady-state precession (FISP)
see steady-state free precession (SSFP)

fast spin echo (FSE)
see turbo spin echo

fat saturation or fat suppression
a technique for eliminating the appearance of fat on an image

fat saturation pulse
occurs when an excitation pulse affects only fat; all of the longitudinal magnetization of the fat is lost, while the rest of the tissues remain unchanged

ferromagnetic
an element that is strongly attracted to a magnetic field and can itself be permanently magnetized, such as iron or cobalt.

field of view (FOV)
area of tissue to be imaged in an MRI scan

flip angle
angle by which the net magnetization vector (*M*) rotates after an RF excitation pulse; the amount of the tip measured in terms of the angle between the original B_0 axis (longitudinal axis) and the angle of precession

flow compensation
the use of extra gradient pulses prior to the signal readout to minimize the artifacts caused by flowing or pulsating blood or CSF

flow effects
motion of material being imaged, particularly flowing blood, resulting in many possible effects in the images; can be understood as being caused by time-of-flight effects or phase shifts that can be acquired by excited spins moving along magnet field gradients

fluid-attenuated inversion recovery (FLAIR)
a special inversion recovery sequence with long TI to eliminate the signal of fluid from the resulting images. The TI of the FLAIR pulse sequence is adjusted to the relaxation time of the tissue that should be suppressed. For fluid suppression, the inversion time (long TI) is set to the zero crossing point of fluid, resulting in the signal being nulled.

Fourier transform (FT)
mathematical technique used to separate the frequency components of an RF signal. Named for French mathematician and physicist, Jean Baptiste Joseph Fourier (1768-1830).

free induction decay (FID)
decay of the amplitude of transient RF signal induced by a 90° RF pulse, although the frequency remains the same; more often, refers to the signal itself

frequency (*f*)
cycles per unit time; usually measured as cycles per second, or hertz (Hz)

frequency encoding
generation of frequency differences along a particular direction of a tissue slice for use in spatial localization of MR signal; a frequency-encoding gradient creates a combination of signals at many different frequencies, with the frequency of each nuclei depending on its location along the gradient direction/within the body

functional MRI (fMRI)
a neuroimaging technique used to study activity in the brain. It shows which structures are active during particular mental operations.

ghosting
image artifact where a shifted copy of the object or "ghost" appears elsewhere in the image. A common cause is gradient distortions in echo-planar imaging or voluntary and involuntary patient motion.

gradient coil
coils that create a magnetic field whose strength is linearly proportional to the distance from the center of the main magnet. An MR scanner has three sets of gradient coils that vary the field in the principle directions of x, y, and z. The rapid switching of the coils accounts for the "banging" noises heard during an MR exam.

gradient echo (GRE)
MR signal that appears following the rephasing of spins by a magnetic gradient in a gradient-echo pulse sequence, which consists of an RF excitation pulse of 90° or less followed by pulses or reversals of magnetic field gradients; unlike spin echo, has no refocusing 180° pulse after the initial excitation

gradient field
a magnetic field with a strength that changes depending on the location within the magnet; generated by gradient coils; also known as just "gradient." Gradient fields add to or subtract from the main magnetic field and are used for slice selection and frequency encoding, altering the MR signal depending on location within the magnet.

gyromagnetic ratio (γ)
the ratio of the magnetic moment to the angular momentum of a particle, which is a constant for a given nucleus; also called magnetogyric ratio. *See also Larmor equation.*

hertz (Hz)
the standard (SI) unit of frequency; equal to the old unit *cycles per second*. Named for German physicist Heinrich Hertz (1857-1894), who made significant scientific contributions to the field of electromagnetism.

inhomogeneity
absence of homogeneity or uniformity; inhomogeneity in a magnetic field occurs when one area of the field deviates from the average magnetic field strength due to the manufacturing process or the presence of air or metal nearby

interleaving
arranging the order of the slice acquisitions so that the slices located next to each other are excited as far apart in time as possible

inversion recovery sequence (IR)
a pulse sequence where an initial 180° pulse is followed by a 90° pulse, resulting in T1-weighted images. This sequence is often used to suppress the signal from a particular tissue such as fat or CSF based on its T1 relaxation time. *See also FLAIR and STIR.*

ionization
the creation of an atom with a net positive or net negative charge due to loss or gain of electrons in the orbits around the nucleus

***k*-space**
the domain in which the information from each phase-encoding step is placed during a pulse sequence. Each "filled in" line of *k*-space corresponds to each phase-encoding step; once the required amount of *k*-space is filled, image reconstruction with a Fourier transform can begin.

Larmor equation
mathematical expression that states that the precessional frequency of a sample of nuclei (such as hydrogen) within an external magnetic field is proportional to the magnetic field and gyromagnetic ratio

$$f = \gamma B_0$$

Named for Irish physicist and mathematician, Joseph Larmor (1857-1942).

Larmor frequency
the frequency at which magnetic resonance is produced in a sample of hydrogen nuclei or other types of nuclei used in MRI; the frequency at which the hydrogen nuclei precess when disturbed from their alignment in the B_0 magnetic field

longitudinal magnetization
component of the net magnetization vector (M) oriented in the same direction as the static magnetic field (B_0)

longitudinal relaxation
restoration of longitudinal magnetization to its equilibrium value; characterized by emission of energy from resonating nuclei; also known as spin-lattice relaxation or T1 relaxation

magnet, permanent
made of materials like magnetized ceramics and capable of producing magnetic fields up to about 0.3T. Permanent magnets are always magnetic and do not require energy to work.

magnet, resistive
uses the physical properties of electricity and magnetism; also called electromagnetic. An electrical current is passed through a loop of wire to generate a magnetic field around the wire. The resistance to the flow of energy through the wire causes the magnets to heat up when in operation, one of the major limitations of this type of magnet.

magnet, superconducting
most commonly used in MR scanners. They also use electricity but at an extremely low temperature so that the current-conducting material loses its resistance for electricity, creating a constant magnetic field. Once the current begins to flow, it can continue almost indefinitely without the need for additional power. However, these magnets must be cooled to near absolute zero with liquid helium or will lose their superconducting properties.

magnetic moment
the net magnetic properties of an object or particle (such as a magnetic dipole)

magnetic susceptibility
changes that distort the normally constant B_0 field arising from material properties, eg, metal implants or air pockets like the paranasal sinuses

magnitude
the amplitude or strength of a vector quantity

matrix
grid of pixels used to construct an MR image; defined by the number of frequency-encoding and phase-encoding steps used in data acquisition

maximum intensity projection (MIP)
a projection image that is obtained from a 3D data set by selecting the maximum intensity along lines or rays that cut through the 3D image volume. Also maximum intensity pixel projection.

nephrogenic systemic fibrosis (NSF)
a rare but potentially serious condition that has been associated with the use of gadolinium-based contrast agents in patients with kidney disease

noise
random and unwanted fluctuations in a signal arising from random motions of particles. Noise causes degradations in the quality of the acquired images.

number of excitations (NEX)
number of image acquisitions per tissue slice that occur during an MRI scan; also known as averaging (AVG) or number of signal averages (NSA)

oblique angle
an angle that is not a multiple of 90° as opposed to an orthogonal angle, which is a right-angle

oversampling
increasing the sampling FOV for the image acquisition while only displaying the smaller field of interest at the prescribed matrix. Used to avoid aliasing.

parallel alignment
in MRI, refers to alignment of spin along the same direction as the static magnetic field B_0

parallel imaging
use of multiple receiver coils to accelerate the acquisition of images, reducing the scan time by a factor of two or more.

paramagnetic
an element that is slightly attracted to a magnetic field, eg, oxygen or gadolinium

perturb
see excitation

phase
particular stage or point of advancement in a cycle; the total number of rotations (or fraction of a rotation) made while the nuclei spin

phase coherence
state in which rotating objects move in phase or unison

phase contrast imaging
an imaging technique that applies extra gradient pulses that are sensitive to moving tissues or flowing blood and can therefore be used to generate angiograms. Compared to time-of-flight imaging, phase contrast imaging typically has lower resolution but can measure the velocity and direction of blood flow.

phase encoding
generation of phase differences along a particular direction of a tissue slice for use in spatial localization of MR signal; a phase-encoding gradient alters the relative position or phase of the hydrogen nuclei as they spin

pixel
smallest discrete part of a digital image (2D) display; from “picture element”

precession
“wobbling” rotation of a spinning object; the spin axis of the precessing object describes a cone-shaped path

proton density-weighted (PD-weighted)
a combination of short TE and long TR where the image depends little on T1 or T2, and the brightness mostly depends on the number of hydrogen protons generating signal at each location

pulse sequence
set of RF magnetic field pulses, gradient waveforms, NMR signal recordings, and the time relationships between them that describes the sequence of steps used to produce MR images

radial imaging
a gradient-echo-based method that has no refocusing pulse and acquires data in a "fan" pattern from the center of *k*-space towards the edge instead of line-by-line. Able to make short T2 tissues visible but requires special reconstruction techniques.

radiography
the use of x-rays to view unseen or difficult-to-image objects. Also referred to as Röentgen rays after Wilhelm Conrad Röentgen (1845-1923), who first described the properties of x-ray

refocusing or rephasing pulse
a 180° RF pulse that flips the direction of precessing hydrogen nuclei. The change in the direction of the magnetization causes the phase of spins to move back together (refocus or rephase) and eventually form an echo.

repetition time (TR)
time interval between two RF excitation pulses in an MRI pulse sequence; also time to recovery or recovery time

resonance
state of a system where a driving force or energy at a preferred oscillation frequency (resonant frequency) can create large changes in the amplitude of oscillations in the system as energy is transferred to the system

respiratory gating
synchronization of imaging with the respiratory cycle to 'freeze' motion from breathing

RF coil
RF coils create the B_1 field that rotates the net magnetization in a pulse sequence. They may also detect the transverse magnetization as it precesses in the x, y plane. Each of these RF coils must resonate, that is, they must efficiently produce and detect energy at the Larmor frequency of the nucleus being examined for the specific field strength of the scanner.

RF pulse
radiofrequency pulse that causes magnetic resonance

saturation
use of an excitation pulse to cause the magnetization of a tissue (such as fat) or region of the body to become zero. This is most frequently used to suppress the tissue from appearing in the acquired image.

short-T1 inversion recovery (STIR)
use of an inversion recovery pulse sequence to eliminate the signal from a tissue with short T1 relaxation time. Most commonly used to suppress signal from fat in the image. Also short-tau inversion recovery.

signal-to-noise ratio (SNR or S/N)
amount of true signal relative to the amount of random background signal (noise) on an image

slice-selection gradient
gradient field that allows excitation and examination of a specific thin slice of tissue

spatial resolution
defines how much detail can be captured in an image and is dependent on the matrix size acquired; the smaller the voxel size, the higher the spatial resolution and the most critical of the three primary requirements of a highly diagnostic MRI exam: spatial resolution, SNR, and image contrast.

Specific Absorption Rate (SAR)
the RF power absorbed per unit of mass of an object, measured in watts per kilogram (W/kg); relates to heating effects of RF pulses

FDA SAR limits:
- 4 W/kg averaged over the whole body for any 15-minute period
- 3 W/kg averaged over the head for any 10-minute period
- 8 W/kg in any gram of tissue in the head or torso for any 5-minute period
- 12 W/kg in any gram of tissue in the extremities for any 5-minute period

spin
the intrinsic angular momentum of an elementary particle(s), like a nucleus, that is also responsible for the observed magnetic moment

spin echo (SE)
MR signal that appears due to the rephasing of spins by a 180° RF refocusing pulse that follows the initial 90° RF pulse in a spin-echo pulse sequence

spin-lattice relaxation time
see T1

spin-spin relaxation time
see T2

steady-state free precession (SSFP)
most closely related to gradient-echo imaging as there is no refocusing excitation pulse. The difference is that all of the gradients used are symmetric, which helps preserve as much of the signal as possible throughout the acquisition. The repetition times used are very short, which makes SSFP less sensitive to motion from breathing, for example. The image contrast is based on a combination of T1, T2, and proton density. Also known as steady-state imaging (GE: FIESTA™; Siemens: FISP™; Philips: Balanced FFE™).

T1
time constant that characterizes the rate of longitudinal relaxation; time for 63% of a tissue's longitudinal magnetization to recover

T1-weighting (T1W)
generation of MR images under conditions that highlight differences in T1 between tissues

T2
time constant that characterizes the rate of transverse relaxation in a perfectly homogeneous magnetic field; time for 63% of a tissue's transverse magnetization to decay

T2-weighting (T2W)
generation of MR images under conditions that highlight differences in T2 between tissues

T2* (T-two-star)
time constant that characterizes the rate of transverse relaxation in an inhomogeneous magnetic field; also characterizes FID

TE
echo delay time; time interval between the initial RF excitation pulse and the echo of a spin-echo or gradient-echo pulse sequence; also echo time

TE/2
time between a 90° and 180° pulse in a spin-echo pulse sequence

tesla (T)
the preferred (SI) unit of magnetic flux density. One tesla equals 10,000 gauss, the older (CGAS) unit. Current range for patient imaging is 0.3 T – 3.0 T. Named for "The Father of Physics," Nicola Tesla (1856-1943) from Croatia, for his contributions to the field of electricity and magnetism.

TI
inversion time; in inversion recovery, the time between the middle of the inversion (180°) RF pulse and middle of the subsequent excitation (90°) pulse to detect the amount of longitudinal magnetization; also time to inversion

time-of-flight (TOF)
a pulse sequence that makes stationary materials appear dark on the image, while moving tissues such as blood show up bright. The resulting image shows only flowing blood, which in turn provides an outline of the blood vessels.

tomographic
imaging by sections or sectioning; cross-sectional images

TR
repetition time; time interval between two RF excitation pulses in an MRI pulse sequence; also time to recovery or recovery time

transverse magnetization
component of the net magnetization vector (*M*) oriented perpendicular to the static magnetic field; the magnetization that can be detected by a receiver coil.

transverse relaxation
decay of transverse magnetization to zero and characterized by spin dephasing; also known as spin-spin relaxation or T2 relaxation

turbo factor
the number of echoes recorded during additional refocusing pulses and that fill in additional lines of *k*-space during a turbo spin-echo pulse sequence

turbo spin echo (TSE)
use of additional refocusing pulses, which generate additional echoes to fill in more lines of *k*-space. The number of echoes recorded during each repetition is called the turbo factor or the echo train length. [TSE- Siemens and Philips; also fast spin echo (FSE-GE)].

vector
describes physical quantities that have both a strength and a direction. A common example would be wind, which has a speed and a direction.

voxel
volume of tissue corresponding to a pixel on an MR image; from “volume element”

wavelength
the distance between the two nearest corresponding points on the wave. Measuring corresponding points between the peaks, the valleys, or any other point yields the same result.

wrap or wrap-around
see aliasing

x, y, z coordinate system
three primary directions to which the three sets of gradient coils are aligned; also called coordinate axes