UNIVERSITY HONORS PROGRAM

SENIOR PROJECT - APPROVAL

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PROJECT TITLE: An Introduction to Magnetic Resonance Imaging Physics

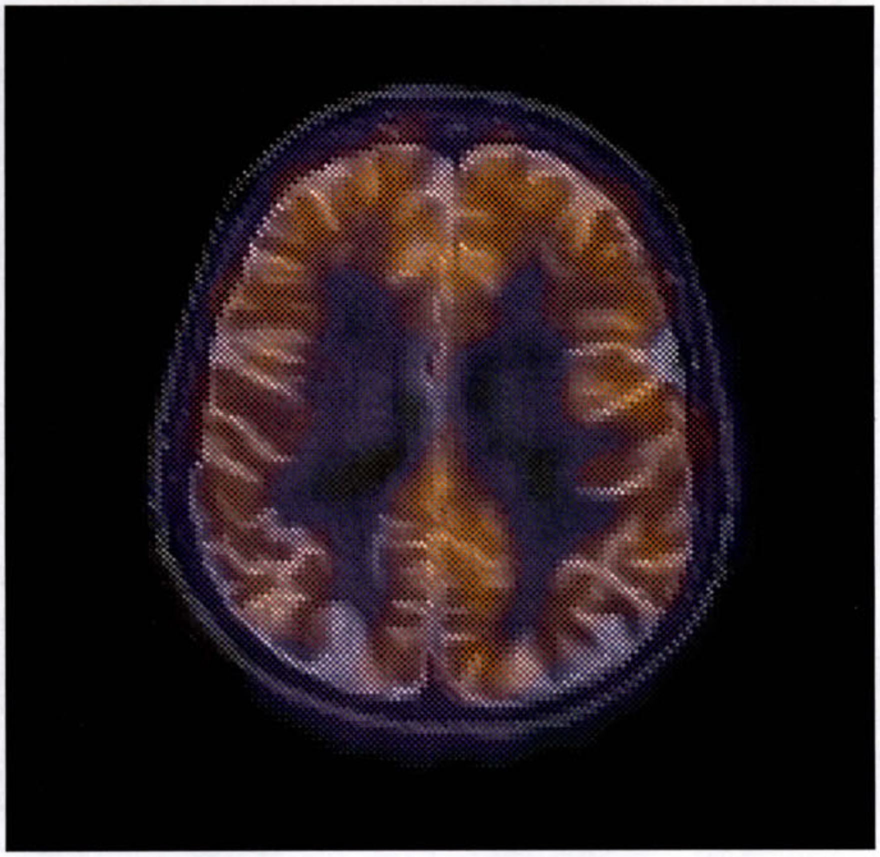
I have reviewed this completed senior honors thesis with this student and certify that it is a project commensurate with honors level undergraduate research in this field.

Signed: Robert L. Toal, Faculty Mentor

Date: 12/14/00

Comments (Optional):

Well Done - very hard worker. It was a pleasure working with Doug this year. He should Do UT proud in the future!



An Introduction to Magnetic Resonance Imaging Physics

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To Dr. Toal, many thanks for his patience and expertise

Abstract

Since its inception in the late 1970s, magnetic resonance imaging (MRI) has been pushing the limits of diagnostic medicine. Diagnoses which were made with great difficulty (or not made at all) using computed tomography or x-ray imagings are now made with relative ease with MRI. One of the most exciting aspects of MRI is the utility it has in so many medical specialties. For example, MRI is currently being used to study the functioning within the brains of Alzheimer's patients, take "snapshots" of the human heart, and trace nerve tracts within the human brain. However, the physical concepts upon which MRI technology was built seem to often be misunderstood or ignored. An understanding of these physics is essential to fully appreciating the images MRI provides us and capabilities MRI will offer clinicians in the future. The purpose of this forum is to provide the reader with a comprehensive overview of MRI physics in a practical, easily readable manner.

Introduction

Magnetic resonance imaging magnets produce very large magnetic fields. An MRI-induced magnetic field of 1.5 Tesla (1 Tesla (MKS units) = 10,000 Gauss (CGS units)) is 30,000 times greater than the strength of the Earth's magnetic field (0.00005 Tesla) and 15 times greater than the magnetic field that exists on the surface of common refrigerator magnets (0.1 Tesla). During the relatively brief history of magnetic resonance imaging (MRI), three different varieties of magnets have been utilized in obtaining images. These magnets are classified as permanent, resistive, and superconducting. However, while both permanent and resistive magnets have their respective advantages, over 95% of the imagers today make use of superconducting magnets (1).

Imaging Magnets

Permanent Magnets. Permanent MRI magnets consist of approximately one hundred tons of bricklike solids made from a ceramic ferromagnetic alloy (1). This type of magnet is comparatively inexpensive to produce, and its low maintenance cost makes permanent MRI magnets a very cost-effective alternative in smaller clinical settings. The two poles of the magnet are situated above and below the patient, producing a vertically oriented main magnetic field (along y-axis), and provide a spacious area for the patient (see Figure 1). This "open" design is advantageous in instances where patients suffer from claustrophobia; therefore, permanent magnets are often used in "open" MRI scanners (10). However, this classification of magnet has its share of shortcomings. Permanent magnets are only capable of producing magnetic fields up to the order of 0.3 Tesla (T) and consequently are not suitable for all diagnostic purposes. The fact that the magnetic fields are fixed and that great difficulty must be taken in producing a uniform magnetic field with these magnets are two other disadvantages associated with permanent magnets (1).

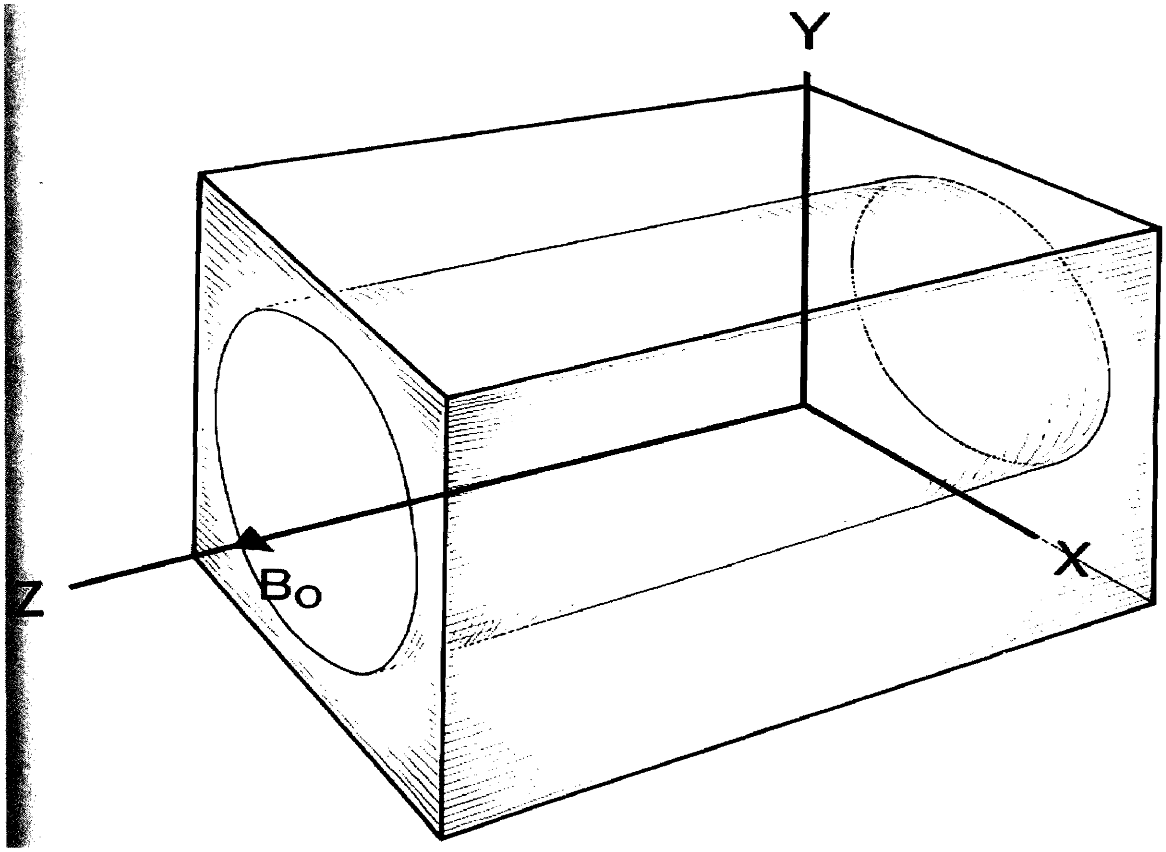

Figure 1: MRI scanner coordinate system (see reference 2)

Resistive Magnets. Resistive MRI magnets (electromagnets), much like permanent magnets, allow for a nonconfining patient area and form a vertically oriented magnetic field (see Figure 2).

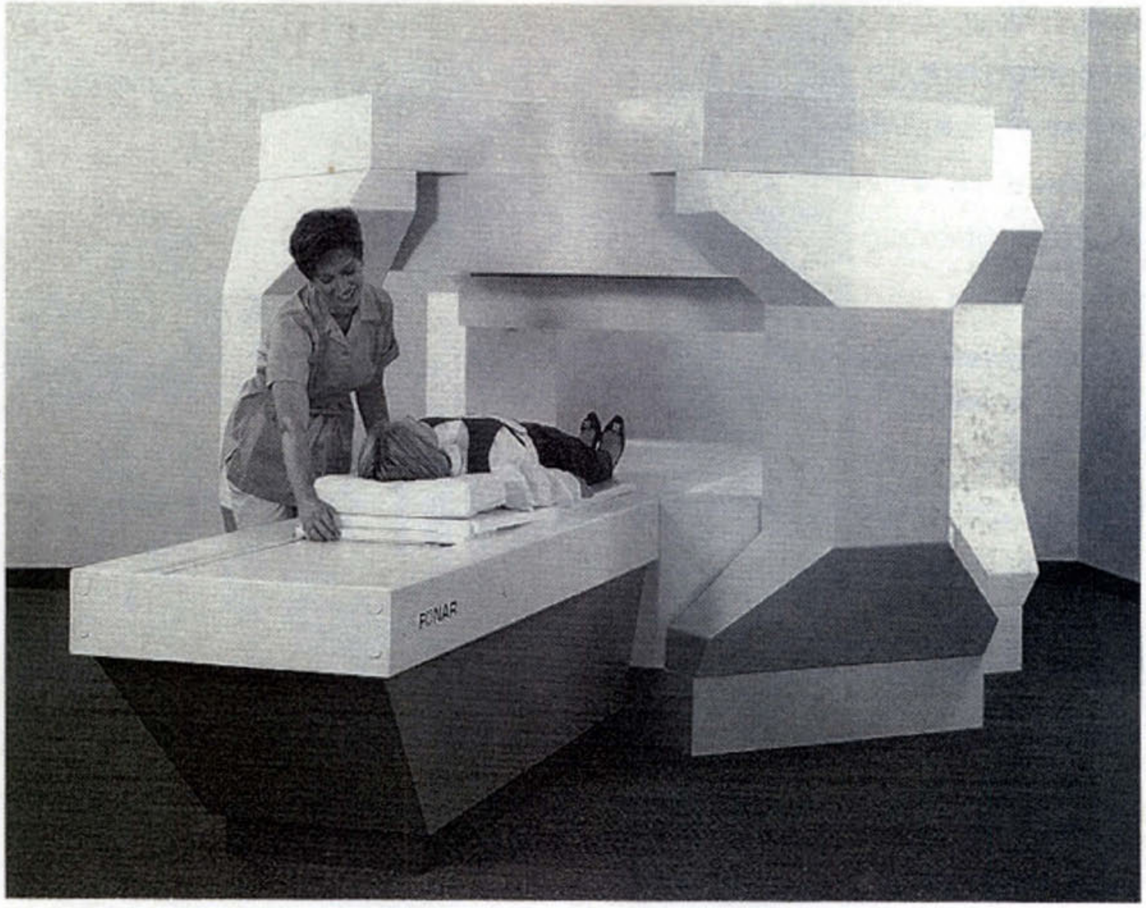

Figure 2: Open MRI scanner (see reference 10)

This main magnetic field is created by sending an electric current through concentrically arranged coils of copper wire, which lie in a plane horizontal to the patient. Magnetic fields up to 0.6 T are produced by the coils and exert large forces on the remaining coils, which make it necessary to rigidly secure each coil.

The magnitude of the magnetic field needed for MRI requires an enormous amount of electrical current. This large power prerequisite can prove to be very expensive! An added consequence of this power necessity is the fact that the coils must be water-cooled. Although only a small amount of resistance retards the current within the copper coils, the enormous amount of electrical current causes the coils to become extremely hot.

Superconducting Magnets. In the early 1960s, a phenomenon was discovered in which certain metal alloys acted as perfect conductors at temperatures near absolute zero (0 degrees Kelvin). Gradually, materials have been discovered that function as superconductors at higher and higher temperatures. Magnets whose function depends upon this discovery are referred to as superconducting magnets and represent over 95% of the magnets used clinically today. The magnetic field is produced by passing current through *miles* of superconducting wire that is precisely wound into several coils that ring the patient (10). Unlike permanent and electromagnets, the magnetic field of superconducting magnets is oriented parallel to patient, along the z-axis (see Figure 1). Recently, materials have been identified that function as superconductors room temperature which promise to have a significant influence on MRI in the near future (8).

The advantages of superconducting magnets should be rather obvious. Since no resistance to electric conduction is present in superconducting coils, heat dissipation is not an issue. An added benefit of the absence of resistance is that once an electrical current is introduced into a superconducting coil, it will remain indefinitely. The ability to produce larger magnetic fields than either permanent or resistive magnets is also an unmistakable advantage. Superconducting magnets are often used to produce fields of the order of 1.5 T or 3.0 T for commercial use; however, larger magnetic fields have been produced for the purpose of research (8). For example, at the Ohio State University, a scanner is in use that can produce B-fields up to 8.0 T in magnitude.

The main disadvantage of superconducting magnets in MRI scanners is the fact that the coils must be maintained at a cryogenic temperature at all times. In order to accomplish this task, manufacturers use a combination of liquid nitrogen and helium chambers called a cyrostat (see Figure 3). The nitrogen chamber acts as a buffer between the external environment and the coils while the helium chamber actually performs the job of supercooling the coils (1). Additional electricity is needed to operate the refrigeration units to keep the cryogens at the proper temperature (8). Due to this cyrostat design, many superconducting devices currently in use do not provide the added patient area of permanent and resistive magnet scanners. However, open MRI devices utilizing superconducting technology have begun to surface (8).

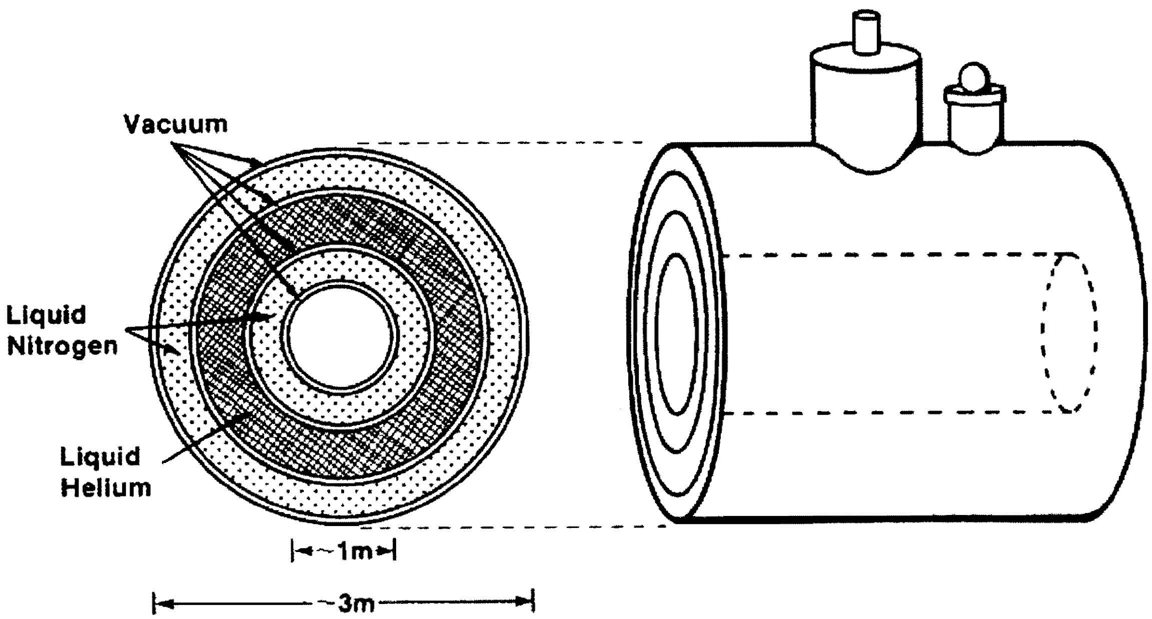

Figure 3: Nitrogen and helium chambers of cyrostat (see reference 1)

Magnetic Resonance Imaging Physics

Spin and Its Consequences. Subatomic particles such as protons and neutrons are known to possess a characteristic spin of either +1/2 or -1/2. Much like particle charge, spin of subatomic particles is an additive property; therefore, by summing spin of individual particles, it is possible to obtain a "net" nuclear spin. Luckily, spin values are not randomly distributed among nuclear particles. In fact, for each pair of like particles (example: proton-proton pair), each entity will possess a spin opposite to its counterpart, and net spin of the pair will be zero. A simple deduction can be made from this observation: net spin of a nucleus with an even atomic number is zero. Thus, many common elements within the body, such as oxygen, carbon, and calcium, possess an even number of nuclear particles and are therefore useless for purposes of MRI (6). However, the most abundant of all tissue components, hydrogen, has an atomic number of only one (remember, a large percentage of the body is water and each water molecule contains two hydrogen atoms) and therefore will possess a net spin equal to the spin of that one proton. **Note**: Since a hydrogen nucleus contains only one proton, the words "hydrogen nucleus" and "proton" will be used interchangeably throughout this text.

Laws of quantum mechanics dictate that a magnetic field will be produced by the spin of a charged particle (neutrons, with a charge of zero, can therefore be ignored). The magnetic field direction is determined by a rule known as the right-hand-rule. Taking your right hand, if you curl your four fingers in the direction of spin and extend your thumb, the direction of your thumb will correspond to the north pole of the magnetic field produced by the particle. The direction of the north pole is often denoted by a vector (see Figure 4) known as a "nuclear" magnetic moment (9).

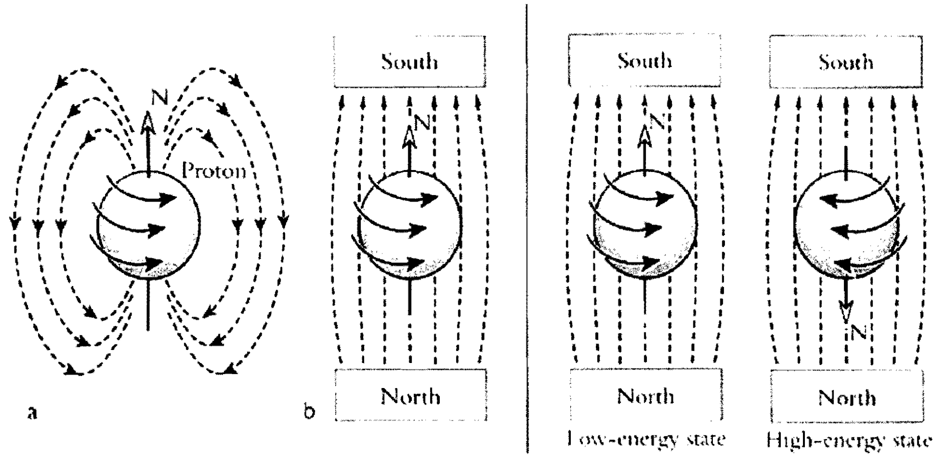

Figure 4: Magnetic field produced by a spinning proton (see reference 10)

In soft tissue of patients, as in most materials, magnetic moments of protons are oriented in a random manner. Since individual magnetic moments of particles is also an additive property, random orientation essentially cancels out the effect of each individual magnetic moment and results in a net magnetization of zero within tissue (1). If, however, the patient is placed in a strong external magnetic field, these small magnetic moments will align either parallel or anti-parallel to the external magnetic field (see Figure 5). The nuclei aligned parallel to the external magnetic field are at a lower energy level than those of the opposite orientation. Intuitively, this should seem natural since the north pole of the particle's magnetic field is aligned with the south pole of the main magnetic field (just as with bar magnets, opposites attract and like poles repel). Due to this advantageous energy state, a slightly larger portion of the spinning nuclei (less than 1 in 10^6 additional protons) are aligned parallel to the external field (B_0) than against it. When this occurs, the patient is said to possess a small net magnetization

(M_0) in the direction of the external magnetic field (see Figure 5). The reason that some protons align anti-parallel is because of thermal molecular interactions induced by molecular motions, rotations, and vibrations that impact proton orientation while in the magnetic field (8).

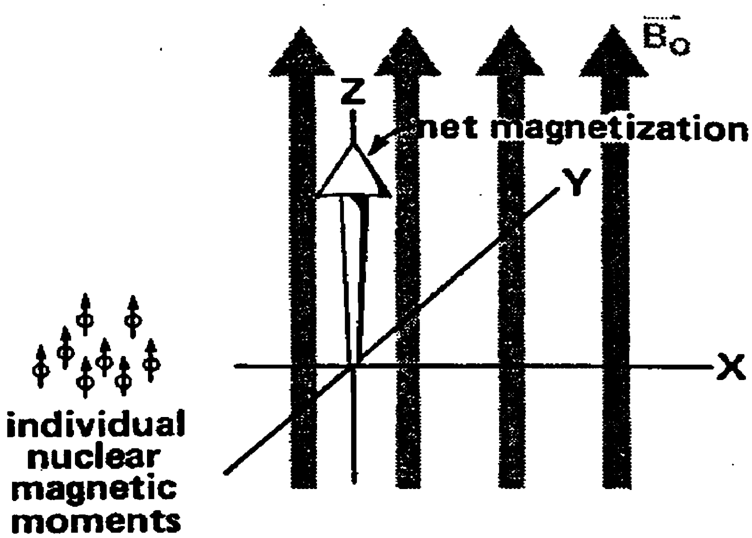

Figure 5: Net magnetization of patient in MRI scanner (see reference 1)

Nuclear Precession. Not only do small particles spin, but in a strong magnetic field, they also have a tendency, much like a gyroscope, to wobble in a circle about the axis of spin (9). This wobbling is known as precession. Thus, as it applies to MRI, nuclei not only spin but also precess (wobble) about the external magnetic field lines (see Figure 6).

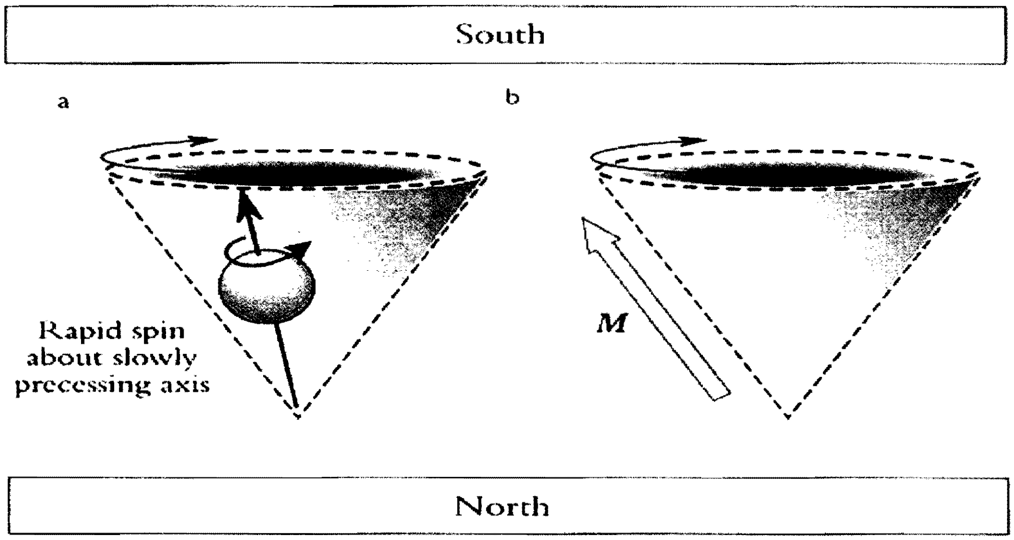

Figure 6: Precession of nuclei (see reference 10)

The frequency with which hydrogen nuclei precess about the external magnetic field lines is known as the Larmor or resonant frequency. The Larmor frequency of individual nuclei is directly dependent on strength of the external magnetic field and a value known as the gyromagnetic constant.

$$\omega = \gamma(B_o)$$

Where ω = Larmor frequency (MHz)

γ = gyromagnetic constant (MHz/Tesla)

B_o = magnetic field strength (Tesla)

The gyromagnetic constant is a nucleus-specific property (hydrogen: γ = 42.57 MHz/T). Therefore, for a given magnetic field strength, different body elements will precess at different frequencies (7). Although hydrogen nuclei all precess at the same frequency in a given external magnetic field, they do not precess in phase with one another (random phase orientation).

For example, the resonant frequencies of a proton in a 0.5 T and 1.5 T B-field are 21.285 MHz and 63.855 MHz, respectively.

Energy can be transferred to nuclei by electromagnetic waves. In clinical settings, these electromagnetic waves are known as RF (radio frequency) pulses. If the B-field strength within the MRI device is known, then the resonant frequency of the hydrogen nuclei within the patient can be easily calculated. The passage through the patient of an RF pulse possessing this Larmor frequency will cause hydrogen nuclei within the patient to absorb a photon of energy and experience resonance (10). This energy absorption allows those nuclei at a lower energy state to "flip" and jump to a higher energy state (align anti-parallel to the external B-field) while at the same instance causing the hydrogen nuclei to precess in phase with one another (see Figure 8).

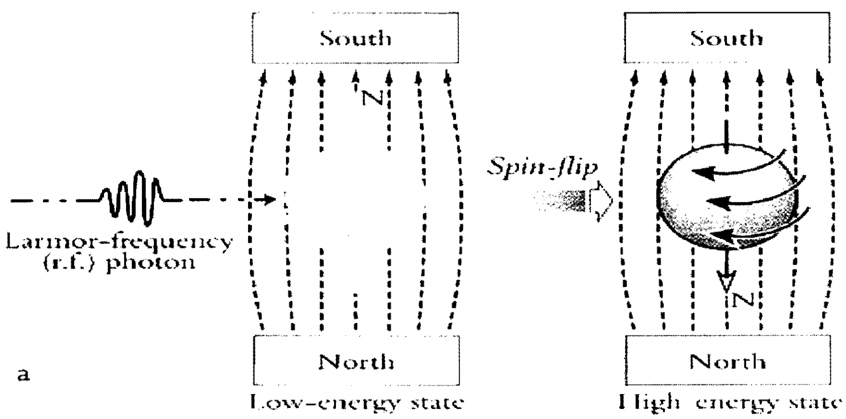

Figure 8: Resonance in hydrogen nuclei (see reference 10)

However, since the electromagnetic wave is sent as a pulse, the excited state of the hydrogen nuclei is only momentary. The nuclei will soon fade back (relax) into their original orientation parallel to the external magnetic field and lose the phase coherence gained by the RF pulse.

Essential Components of MRI Scanners (see Figure 9)

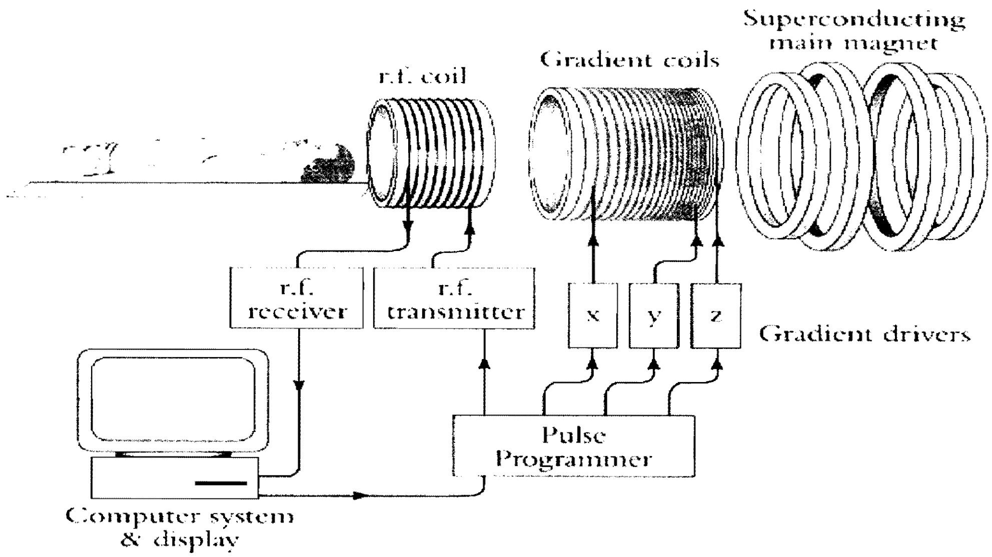

Figure 9: Essentials within an MRI scanner (see reference 10)

Main magnet. The main magnet is the component of the scanner that is responsible for producing a strong main magnetic field. As mentioned earlier, these magnets can be classified into three groups: permanent, resistive (electromagnets), and superconducting magnets.

Shim Coils. Homogeneity of the main external B-field is of utmost importance in MRI scanners. Variation within this B-field will result in the degradation of image quality (10). For example, a 1.0 T magnetic field should vary no more than ±50 μT or ±50 ppm (parts per million). Sufficient homogeneity can be achieved by the use of shim coils. Located just inside the main magnet, each coil possesses its own power supply and is capable of adjusting its current and polarity in such a way that the homogeneity of the main magnetic field is maximized.

Gradient coils. Located just inside the shim coils are three independent coils known as gradient coils (one for each of the x, y, and z-axes). These coils are used in choosing location of the imaging plane within the patient (see *Slice mechanics* for more details). These fields are intermittently switched on and off throughout a scan and are responsible for the muffled, booming sounds that can be heard by the patient (10).

RF/Receiving coils. RF/receiving coils serve a dual purpose in MRI scanners. Their primary task is to produce the RF (radio frequency) pulses that cause protons to absorb photons of energy and therefore result in proton spin flips. However, RF/receiving coils also detect the weak RF signals that are given off by the relaxing hydrogen nuclei. The coils then amplify the signal and send it to a computer for analysis (1).

RF/receiving coils are available in a variety of different shapes, with each shape lending itself to a particular type of imaging (10). For example, an RF/receiving coil known as a saddle coil is used primarily for neuroimaging (see Figure 10, top left corner).

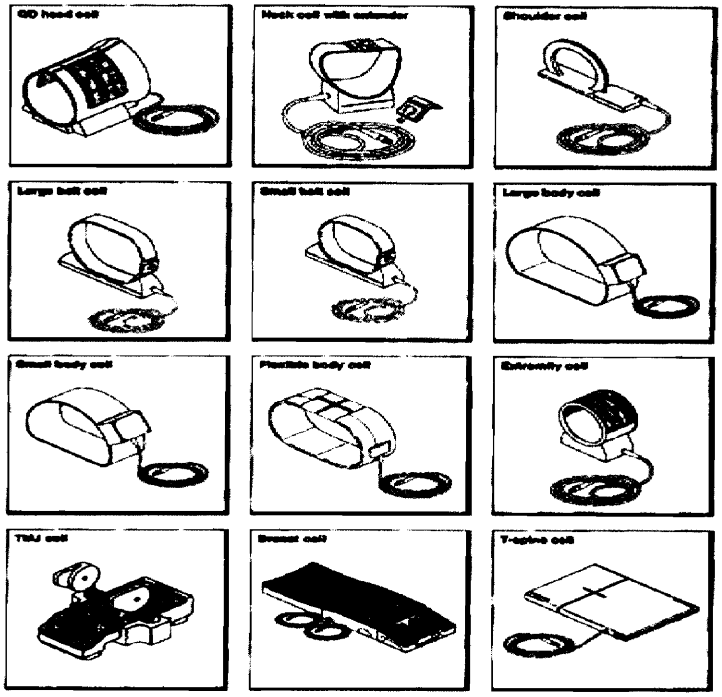

FIG. 7-1. Various coils used in vertical field magnets.

Figure 10: Various RF/receiving coils (see reference 4)

Computer system. The control of gradient coil and RF pulse initiation is under computer control. A computer system is also responsible for storing and analyzing resonant data, carrying out reconstruction calculations, and producing images for display (10).

Magnetic Resonance Imaging Parameters

Computed tomography (CT) and x-ray imaging both use a parameter known as linear x-ray attenuation coefficient to form images. However, the formation of magnetic resonance images is dependent upon three factors: spin density, T1 relaxation time, and T2 relaxation time (7). This accounts for the superior contrast resolution of MRI versus CT.

Spin density. Spin density is a measure of the concentration of hydrogen nuclei within a particular region of interest. The greater the number of hydrogen atoms present, the stronger (brighter) the signal received from the patient. Tissues such as muscle and fat, which are rich in hydrogen atoms, will appear much brighter than materials with little or no hydrogen atom concentration. This is the reason that bone and air appear black in MRI. Since spin density is directly proportional to the number of hydrogen nuclei (protons), it is also known as proton or hydrogen density (1).

T1 relaxation time. When an RF pulse with a frequency corresponding to the Larmor frequency of hydrogen irradiates a patient, some of the magnetic moments (spins) aligned parallel to the strong external magnetic field will absorb energy and flip to an orientation anti-parallel to the external field (see Figure 8). The number of magnetic moments that absorb energy is dependent on the strength and duration of the RF pulse. Therefore, manipulating the RF pulse can control the number of spins that are flipped (7).

As mentioned earlier, the application of an external magnetic field results in a small surplus of magnetic moments parallel to the external B-field. This surplus of spins yields a net magnetization in the direction of this B-field (see Figure 5); in other words, the patient becomes "magnetized." Let's imagine for a moment that an RF pulse has a strength and duration such that all of the spins in this surplus flip to an anti-parallel orientation. As a result of this pulse, the net magnetization of the patient would also flip against the field (see Figure 11). This type of pulse is referred to as a 180° pulse since the net magnetization of the patient is reversed (1).

Let us now consider an RF pulse that flips the magnetic moments of half of the "surplus" hydrogen nuclei. Since this pulse would result in only half of the spins being flipped, the net magnetic moment of the patient would rotate only halfway to the anti-parallel position. In other words, the net magnetization would lie in a plane perpendicular to the plane of the external magnetic field. This type of pulse is referred to as a 90° pulse since the net magnetic moment rotates through a right angle (1).

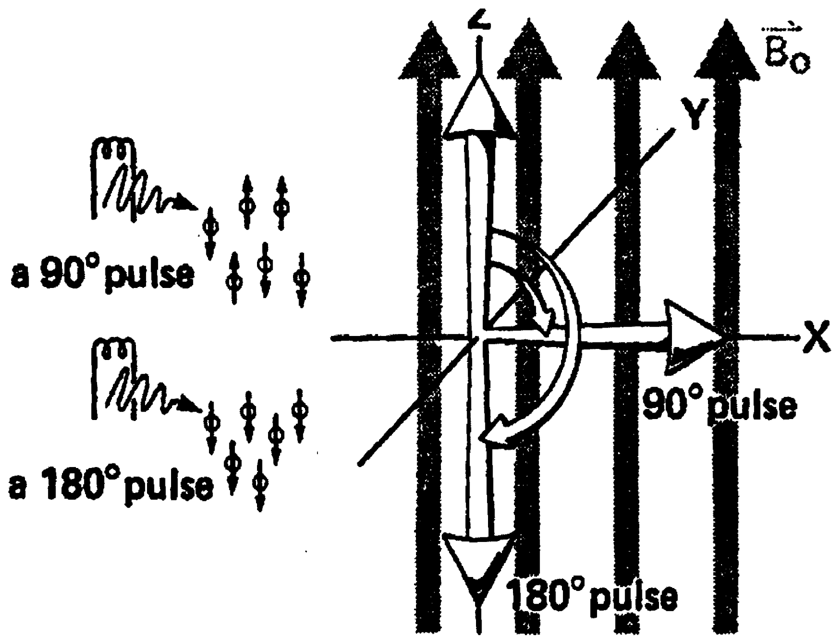

Figure 11: 90 and 180 degree pulses (see reference 1)

Once an RF pulse is terminated, the energized protons begin losing this absorbed energy and fade back (referred to as relaxation) to their original (parallel) position. The released energy is given off in the form of an RF signal (9). A receiving coil positioned parallel to the axis of the external B-field detects these RF signals radiating from these protons. These electromagnetic waves induce a voltage in the coil which is analyzed and used to form magnetic resonance images.

The signal released by the protons is at its strongest when the entire net magnetic moment of the patient is perpendicular (transverse) to the external B-field. As this net magnetization regresses back to a parallel orientation, its transverse component and the signal detected by the receiving coil becomes smaller. This loss of energy and subsequent realignment with the external B-field is known as T1 relaxation (10). T1 relaxation (regrowth of net magnetization) can be defined by the function:

$$1-e^{-\frac{t}{T1}}$$

where t is time and T1 is T1 relaxation time. When t is equal to T1, this equation reduces to 1 - e^{-1}, which has a value of 0.63212. In physical terms, Tl relaxation time is the timed required for the net magnetization of the patient along the z-axis to regain 63.212% of its original value (see Figure 12).

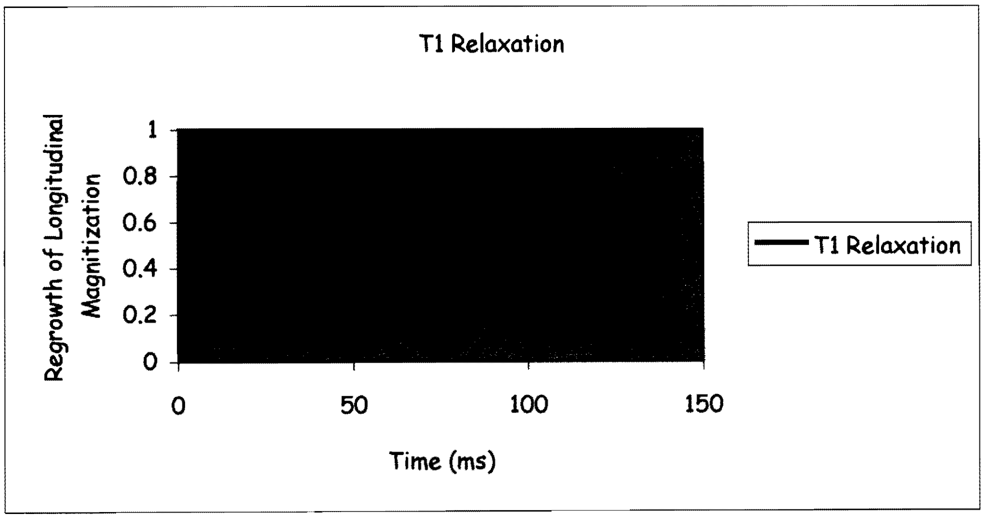

Figure 12: T1 relaxation with T1 relaxation time = 37ms

The return of net magnetization in the longitudinal plane does not give off a signal. This means that T1 cannot be measured directly. Rather, successive RF pulses are used to flip the nuclei from lower to a higher energy state. The difference in the transverse signal given off by the nuclei following each RF pulse is proportional to the number of atoms returning to equilibrium (relaxation). This then becomes a measure of T1 (8).

Each tissue has its own characteristic T1 relaxation time due to the unique manner by which it is held together and the unique magnetic environments these tissues produce. Since this type of relaxation occurs in the plane of the external B-field, T1 relaxation time is also known as longitudinal relaxation time. An additional moniker for this parameter, spin-lattice relaxation time, is due to the interaction of released energy with the lattice framework of the tissue (1).

T2 relaxation time. In addition to flipping the magnetic moments of protons, properly tuned RF pulses also cause protons to precess in phase with one another. Therefore, for a 90° pulse, the net magnetization can be imagined to be lying in the transverse plane and precessing about the axis of the external magnetic field. However, when the RF pulse is terminated, the interaction of individual magnetic moments (fields) causes some protons to precess faster, some slower. This process is known as dephasing (7).

As the protons fall further and further out of phase, the transverse component of the patient's net magnetic moment becomes smaller and smaller. Eventually, the individual magnetic moments will dephase completely and the transverse component will be nonexistent, as it was before the introduction of the RF pulse (see Figure 13).

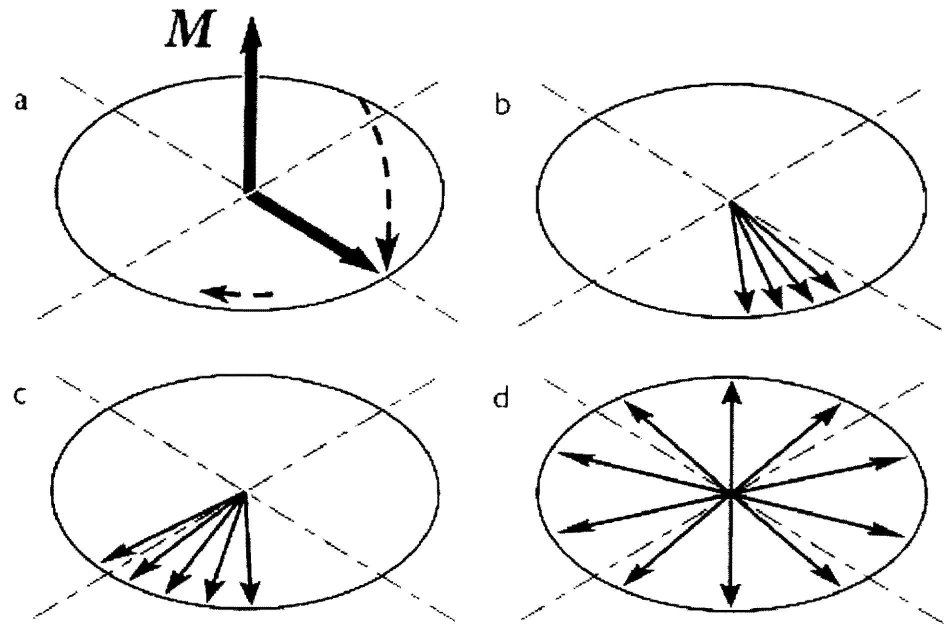

Figure 13: Dephasing of proton spins (see reference 10)

This process of dephasing and subsequent shrinking of the transverse component of the net magnetic moment is known as T2 relaxation. Since this decay of phase coherence results from the interaction of individual spins and occurs in a plane transverse to the external B-field, this parameter is also known as spin-spin or transverse relaxation time (1).

Transverse relaxation with respect to time can be represented by the equation

$$e^{-\frac{t}{T2}}$$

where t is time and T2 is T2 relaxation time (see Figure 14). When t is equal to T2, the equation reduces to e^{-1}, which has a value of 0.36788. In physical terms, T2 relaxation time is the time required for the transverse net magnetization of the patient to fall by 36.788% or to a value that is 63.212% of its original value.

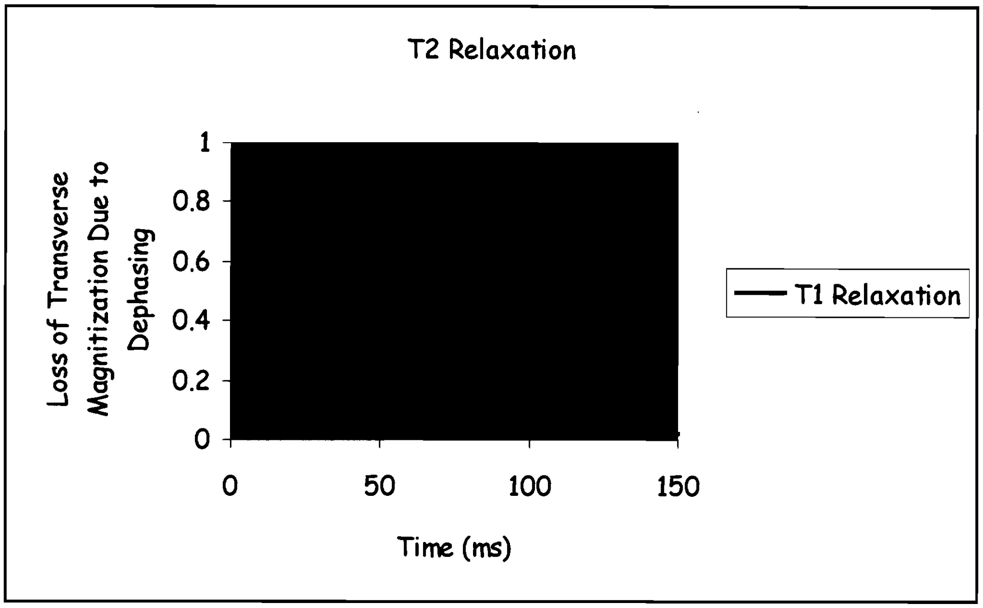

Figure 14: T2 relaxation with T2 relaxation time = 37 ms

Just as with T1 relaxation times, T2 relaxation times differ from tissue to tissue due to their different magnetic environments. In fact, T1 and T2 relaxation times can differ by 20% to 30% between various soft tissue types. This superior contrast resolution is one major advantage that MRI has over both computed tomography (CT) and x-ray imaging (1). X-ray imaging uses linear x-ray attenuation coefficient to form its images. However, this imaging parameter differs by only approximately 1% in soft tissues. These x-ray images are also degraded by scatter radiation. CT imaging also uses linear x-ray attenuation coefficient as its imaging parameter. CT does however eliminate scatter radiation degradation, which provides greater contrast resolution than x-ray imaging.

Slice Mechanics

Slice selection. One distinctive advantage that magnetic resonance imaging has over other imaging modalities is its ability to perform multiplanar imaging without repositioning the patient (10). This method of slice selection involves the use of a group of coils oriented along the x, y, and z-axes of the patient known as gradient coils (see Figure 1). Initiation of a gradient coil forms a magnetic field that can be superimposed onto the main B-field. This superposition of B-fields creates a gradient magnetic field in the direction of the gradient coil's orientation that varies linearly from one extreme of the patient to the other (see Figure 15).

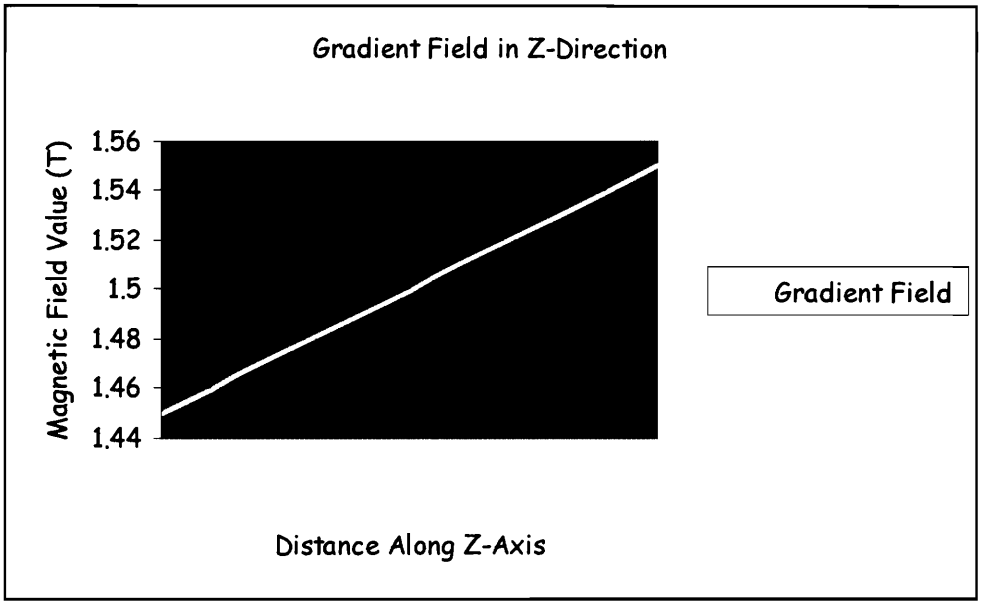

Figure 15: Gradient magnetic field along z-axis with main field = 1.5 T

For example, suppose for a moment that the gradient coil oriented along the z-axis was initiated. The combination of the resulting B-field with the main magnetic field produced a gradient field with values of 1.3 T and 1.5 T at the lower and upper extremities, respectively, of the patient. Since this gradient field is known to vary linearly along the z-axis, the value of the field at any point along the z-axis can be calculated.

You may be wondering how a linearly varying gradient field would be helpful in slice selection. If you recall, the resonant (Larmor) frequency of hydrogen nuclei is dependent upon the gyromagnetic constant of hydrogen and the magnitude of the magnetic field to which a hydrogen nucleus is exposed. Therefore, since the magnetic field at each point along the patient is slightly different, if the B-field magnitude at the plane of interest is known, the appropriate RF pulse can easily be obtained that will excite only those protons within that plane (2). Referring back to the previous example, imagine that physicians desire to image a plane exactly halfway between the lower and upper extremities of the patient. Using a B-field of 1.4 T in the Larmor equation, the frequency needed to induce resonance in that plane could be found. The type of image in the example would be known as a transverse (trans-axial) image. Images obtained from a gradient field along the x-axis are known as sagittal images while those obtained from gradients along the y-axis are known as dorsal (coronal) images. It is also possible to obtain oblique images by using combinations of the x, y, and z-gradients (see Figure 16).

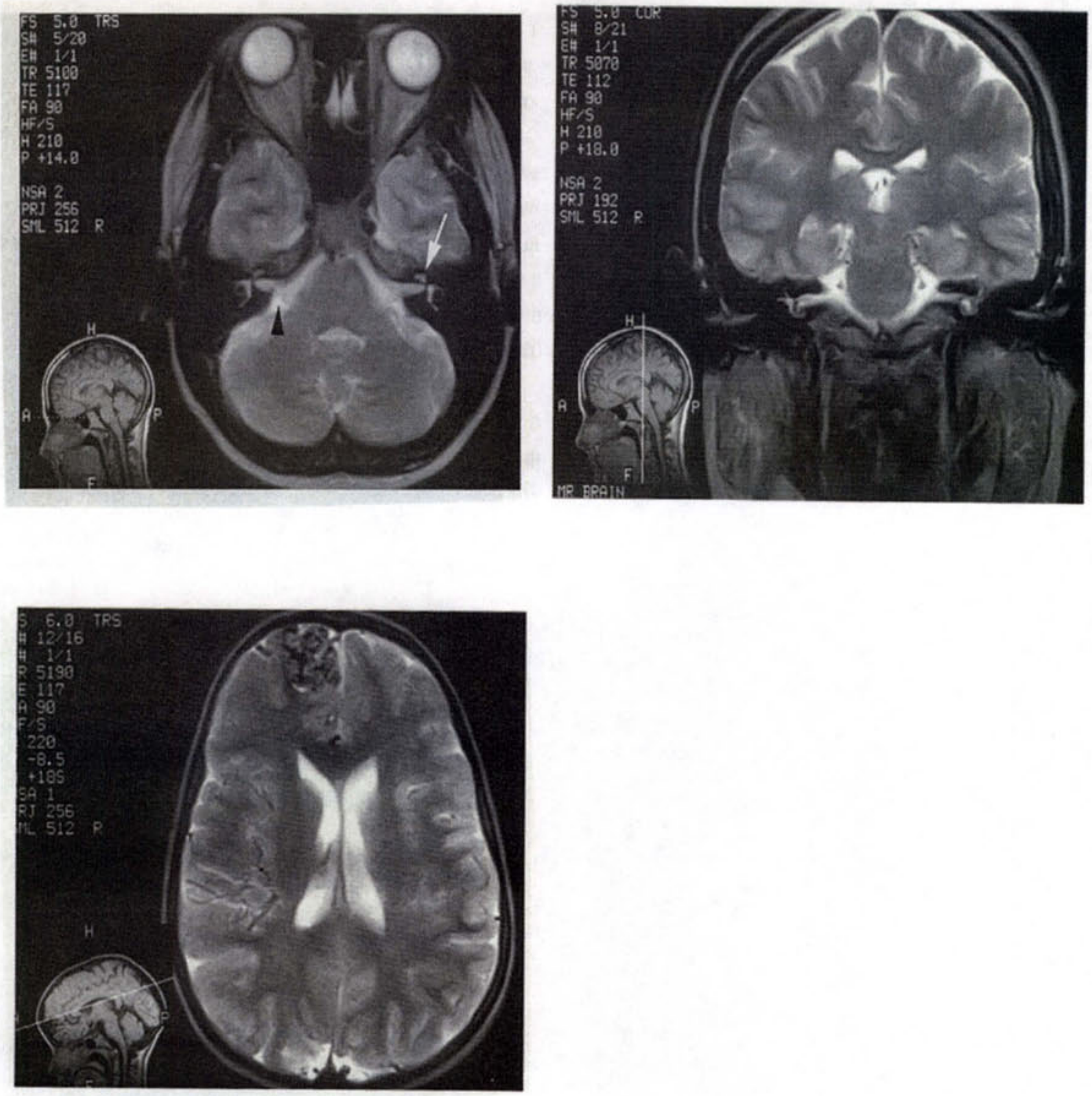

Figure 16: (top left) transverse image; (top right) coronal image; (bottom) oblique image; sagittal images and slice plane can be seen in bottom left corner of each image (see reference 4)

Slice thickness. The thickness of the slices is determined by the range of frequencies included within the RF pulse (this has not been mentioned earlier, but it is possible to send pulses containing a band of frequencies). For example, an RF pulse containing all frequencies from 64 MHz to 64.5 MHz would excite all protons with resonant frequencies within that range.

Slice thickness can also be influenced by the severity of the slope of the gradient field: the steeper the slope, the thinner the slice (2).

Image reconstruction. The formation and timing of gradient fields and RF pulses is under computer control (see Figure 9). A computer also has the responsibility of analyzing data from the receiving coil, carrying out reconstruction calculations, and producing the resulting images for viewing. The mathematical procedure that is utilized by the computer in the reconstruction calculations is known as the Fourier transform (7). A two-dimensional (2-D) Fourier transform is used in the analysis of slices formed by a single-frequency RF pulse (a two-dimensional plane). In order to determine the fraction of the signal that originated from each portion of the plane, the slice must be divided into tiny squares, similar to the divisions present on graphing paper. These tiny squares are known as pixels and this process is known as spatial localization (2). Spatial localization is achieved with the gradient coils within the scanner.

Once the slice has been selected by the slice-select gradient, the remaining two gradient coils are used to divide the slice into strips and then into squares. The remaining two gradient coils are referred to as the frequency-encoding gradient and the phase-encoding gradient (3). In order to grasp a better understanding of this process, let us again refer back to the example of the transverse slice. In this particular case, the gradient coil along the z-axis would be the slice-select gradient, and the x and y-axis gradients would be the frequency-encoding and phase-encoding gradients, respectively (with the x-axis gradient dividing the plane into vertical strips and the y-axis gradient dividing it into vertical strips). Once the slice has been spatially encoded, the Fourier transform (in very simplified terms) can convert the data in each pixel into an intensity or gray-scale value. The compilation of these intensity values, which range from white (highest intensity) to black (lowest intensity), result in a viewable image.

A three-dimensional (3-D) Fourier transform is used in the analysis of slices formed through the application of an RF pulse containing a range of frequencies (2). The plane of interest is possesses a finite thickness and can be considered as a volume element. Appropriately, the divisions into which the volume element is separated are known as voxels. As one might expect, the greater the number of divisions within the slice, the more precise the image reconstruction will be for each point in space and the greater the spatial resolution in the resulting image (this is applicable for the 2-D case as well).

Hazards of MRI Scanning

Numerous studies have been performed on the effect of strong magnetic fields on biological tissue, and no evidence has ever been found to suggest that MRI scanning is directly harmful to the patient. Further, researchers are extremely confident that no long-term effects will be found by subsequent studies (1). There are, however, indirect dangers associated with the fields produced during MRI scanning.

Cardiac pacemakers. Cardiac pacemakers use a small electromagnet in their operation and exposure to a strong magnetic field can cause the pacemaker to fail. Therefore, those individuals with cardiac pacemakers should refrain from being near MRI scanners and should never be imaged (1).

Other internal metallic objects. Metallic objects within the body such as aneurysm clips, surgical pins, shrapnel, artificial heart valves, etc. often pose no threat to the patient (10). It is, however, advisable for one to inform their physician of any possible source of metallic material within their body to ensure no possible side effects will occur.

Personal Items. Items such as credit cards or wristwatches should not be brought into contact with the strong magnetic fields of MRI scanners. Credit cards contain magnetic strips and can be deactivated by a strong B-field. Tiny moving parts within a mechanical watch can become magnetized and cause the watch to malfunction (1).

Metallic objects. The main magnetic field of a MRI scanner can extent huge forces on magnetic materials made of steel, iron, or other ferromagnetic (easily magnetized) materials. Pens have been known to be sucked right out of the pockets of researchers. Personnel must also be careful when maintenance work is being performed near a scanner. For example, one story surfaced at a prominent medical school in which a maintenance worker veered a bit too close to a superconducting MRI scanner while vacuuming. The vacuum was sucked into the core of the magnet, and the magnet had to be quenched and the cooling mechanism deactivated in order to remove the vacuum cleaner.

Fringe fields. An added problem with resistive magnets is the fact that the magnetic fields produced extend beyond the MRI scanner (see Figure 17).

This excess field is known as a "fringe" field. Fringe fields affect the performance of many types of equipment, most notably computers, electron microscope, and cathode ray tubes, and must be accounted for in the hospital or lab setting.

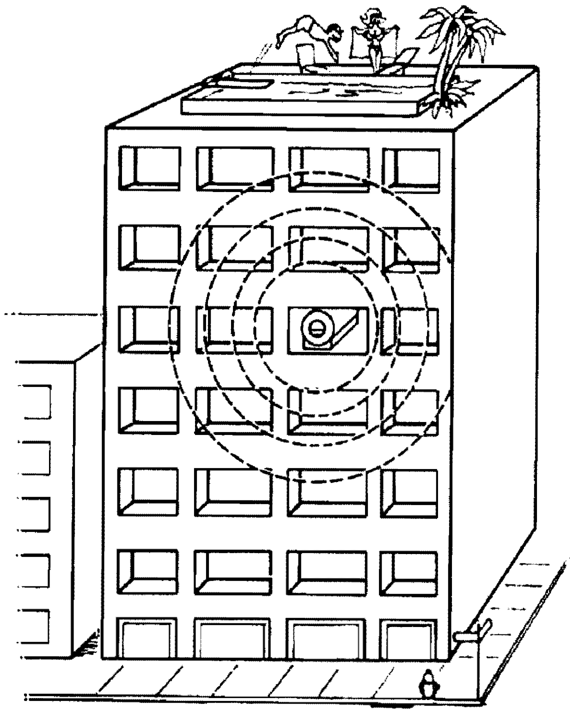

Figure 17: Fringe field due to MRI scanner (see Reference 1)

Conclusion

The discussion thus far has entailed the very basics upon which magnetic resonance imaging is built as well as some practical aspects within MRI. Other subjects, such as RF pulse sequences and the weighting of images to emphasize one parameter over another, is most definitely worthy of consideration but are not within the focus of this forum. MRI has made an incredible impact on the field of diagnostic medicine, and with alternate techniques such as magnetic resonance angiography becoming popular in clinical settings, it promises to become the standard upon which all other diagnostic tools are measured. A thorough understanding of the basic concepts of MRI is essential to fully appreciating the images produced by MRI and the capabilities that MRI will provide to medical professionals in the future.

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