MRI for Technologists

MR Image Formation and Imaging Techniques

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.

MR Image Formation and Imaging Technique introduces the learner to the fundamental pulse sequences used for the majority of MRI acquisitions, the operator-controlled parameters that affect image quality, how images can be enhanced by the use of gadolinium-based contrast agent, how an image is generated, and how the interplay of spatial resolution, signal-to-noise ratio, and image contrast affects image quality.

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

- Describe the two fundamental pulse sequence categories
- Explain the time parameters TE and TR
- Discuss how the flip angle affects image appearance
- Describe how images can be enhanced with the use of contrast agents
- Describe how 2D and 3D images are generated using gradient magnetic fields
- Explain the concepts of *k*-space and Fourier transform
- List several additional imaging techniques and their contrast properties
- Describe how these techniques differ from conventional spin echo and gradient echo pulse sequences
- Explain characteristics that influence image quality
- Describe physical and image acquisition parameters

EDUCATIONAL CREDIT

This program has been approved by the American Society of Radiologic Technologists (ASRT) for 3.75 hours ARRT Category A continuing education credit.

HOW TO RECEIVE CREDIT

Estimated time to complete this activity is 3.75 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

Daniel R. Thedens received his doctorate in electrical engineering from Stanford University. In addition to his research and teaching responsibilities at the University of Iowa, Dr. Thedens is an Associate Research Scientist in Department of Radiology, Division of Diagnostic Radiology - Physics, at the University of Iowa Health Center. He also serves as co-chair for the Radiology MR Research Advisory Board as well as Technical Director of the Small Animal MRI Facility.

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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MRI Image Formation and Imaging Techniques

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

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

- Describe the two fundamental pulse sequence categories
- Explain the time parameters TE and TR
- Discuss how the flip angle affects image appearance
- Describe how images can be enhanced with the use of contrast agents
- Describe how 2D and 3D images are generated using gradient magnetic fields
- Explain the concepts of *k*-space and Fourier transform
- Adjust spatial resolution and image contrast to optimize signal-to-noise ratio and produce diagnostic images
- List several additional imaging techniques and their contrast properties
- Describe how these techniques differ from conventional spin-echo and gradient-echo pulse sequences
- Explain characteristics that influence image quality
- Adjust common types of operator-controlled parameters

POINTS for PRACTICE

1. **Why does a 180° excitation pulse follow the 90° excitation pulse in spin-echo?**
2. **Define TE and TR, two important time constants associated with spin-echo pulse sequences.**
3. **How do TE and TR influence the appearance of tissues in spin-echo pulse sequences?**
4. **What are T1 and T2 weighting?**
5. **The scan operator may choose parameters for T1, T2 and proton-density weighted images. What is a general rule for making these selections?**
6. **TR and TE are two important time constants also associated with gradient-echo pulse sequences. Explain why. What is the role of the flip angle in a gradient-echo pulse sequence?**
7. **What particular kind of flip angle does the Ernst angle describe?**
8. **Explain how gadolinium-based contrast agents interact with water molecules.**

IMAGE CONTRAST

As a preface to this unit on image formation and imaging techniques, a review of the basic principles underlying the generation of signal used to form MR images of tissues is in order. The following are required elements for generating signal:

- A strong magnetic field aligns the orientation of hydrogen nuclei along the same direction as the applied magnetic field. A small majority of these nuclei (a few parts per million) orient in the preferred parallel direction, creating net magnetization *M* in the same direction as the main magnetic field, B_0.
- A second magnetic field, B_1, oscillating at the resonant or **Larmor frequency**, is turned on to reorient or "flip" the spins away from the direction of the main magnetic field, B_0. This is RF (radiofrequency) excitation.
- The "flipped" spins rotate or precess around the direction of the main B_0 magnetic field. This precession can be detected as a voltage in a receiving coil, producing the **nuclear magnetic resonance** (NMR) signal or free induction decay. The signal is recorded and further processed to form MR images.
- Immediately after excitation, both T1 and T2 relaxation begin, resulting in the decay of transverse magnetization (T2 relaxation) and the restoration of longitudinal magnetization (T1 relaxation).

The rate at which the magnetization in different tissues is lost in the transverse plane and is restored in the longitudinal direction affects the signal intensity and appearance of the tissue on the final MR image, resulting in the contrast between different tissue types. In MRI, the steps of excitation and signal recording are repeated many times to form a complete image, and the timing of these steps affects the image appearance.

We will describe the steps used in routine MR image acquisition, demonstrating how T1 and T2 relaxation times and variable scanning parameters influence the appearance of tissues on MR images. Pulse sequence selection plays an important role in image acquisition and appearance, especially for the most commonly used types of sequences: spin-echo and gradient-echo.

Pulse Sequences

The set of RF excitation pulses and signal readout steps used to acquire an MR image form a **pulse sequence**. The two general categories of pulse sequences used for the majority of MRI acquisitions are **spin-echo** (SE) and **gradient-echo** (GRE). Most other pulse sequences can be classified as either being spin-echo-based or gradient-echo-based.

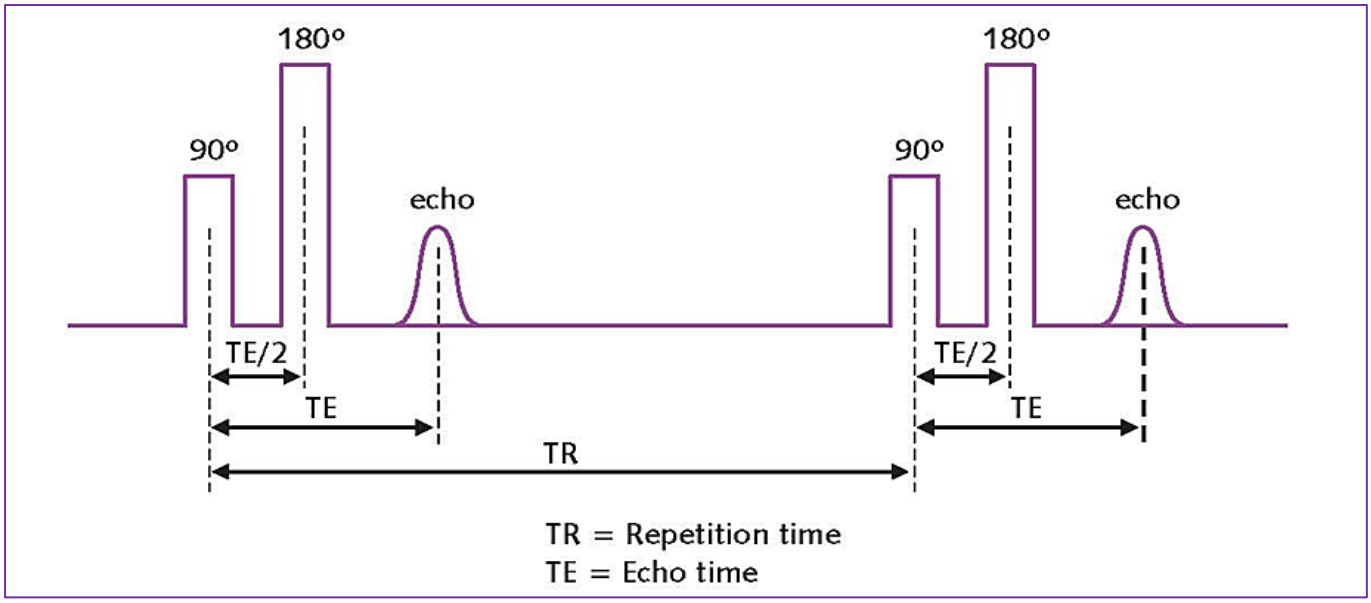

Figure 1. Spin-echo pulse sequence.

Spin-echo Pulse Sequence

90° and 180° Pulses

The spin-echo pulse sequence begins with an excitation pulse of a 90° **flip angle**. After a delay, another excitation pulse of 180° is applied. Finally, after another short delay, the signal arising from the remaining transverse magnetization is recorded. As shown in **Figure 1**, these steps are repeated as many times as needed to generate a complete set of images.

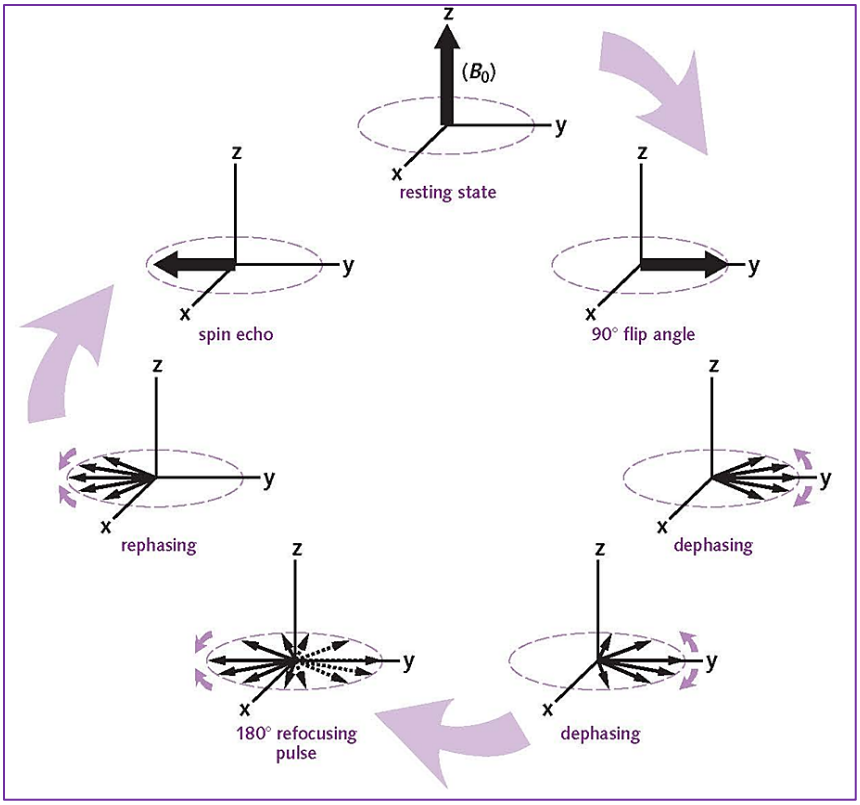

Figure 2. After a 90° excitation pulse, a 180° pulse rephases transverse magnetization, leading to the formation of an echo.

The purpose of the 90° excitation pulse is to flip the magnetization completely into the transverse direction to generate the maximum amount of signal that can be detected.

The 180° pulse that follows is a **rephasing** or **refocusing pulse**. The purpose of the 180° pulse is to flip the orientation of the magnetization around so the spins that have rotated ahead of the others are now lagging by the same amount. Likewise, the spins that have fallen behind are now ahead of the rest (**Figure 2**).

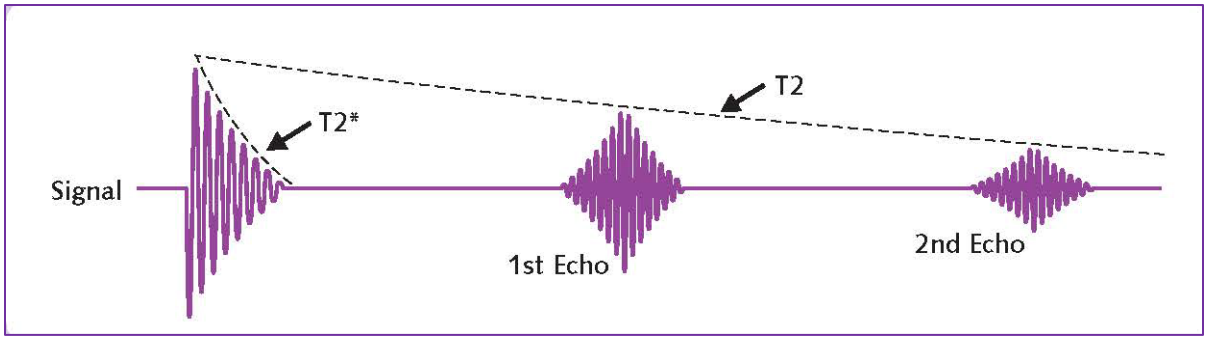

Figure 3. Spin-echo signals are generated by transmitting a 90° excitation pulse, followed by one or more 180° excitation pulses. This refocuses the signal loss from T2*and generates a series of echoes.

There are two factors that cause signal loss — **T2** and **T2*** (T-two-star) relaxation — with the effect of T2* causing more rapid loss of signal. Recall that the cause of this signal loss is slight variation in the main magnetic field, making the direction of the spins "fan out" or **dephase**. Signal from different spins will cancel each other out, resulting in a rapid loss of total signal.

After the 180° pulse reorients the spins, the dephasing or "fanning out" process begins to reverse itself, with the spins beginning to come back towards each other until they are all oriented in the same direction as they were after the 90° pulse. The net result is that all of the signal loss from dephasing due to T2* is eliminated, and the signal is only affected by T2 decay.

The signal fades out due to T2* relaxation, then fades back in as this process is reversed to form a spin-echo (**Figure 3**). Recall that the duration of T2 is usually much longer than the duration of T2*. The spin-echo helps preserve the maximum amount of signal during image acquisition, making the signal depend only on the tissue characteristics and not on the main magnetic field.

Variable Parameters in Spin-Echo Imaging

As noted in Figure 1, there are two important time parameters associated with the spin-echo pulse sequence: echo time (TE) and repetition time (TR). These parameters are controlled by the scan operator and strongly influence the appearance of tissues within the images.

Echo time

The **echo time** or TE measures the time between the initial 90° pulse and the time at which the signal is recorded. It is during this time that the spin-echo forms. Its duration is typically from a few milliseconds to 100 or more milliseconds.

TE/2

The total echo time is divided into two halves of duration — TE/2 — or the time between the 90° and 180° pulses. As seen on the pulse sequence diagram in Figure 1, there is a delay of TE/2 between the initial 90° excitation pulse and the 180° refocusing pulse. During this delay, both T2 and T2* signal loss occur. Another delay follows between the 180° pulse and the signal readout time. During this delay, T2 signal loss occurs, but the T2* effects are refocused so that at the time the signal is recorded, the signal depends only on the T2 relaxation that has occurred. This is desirable, since T2 relaxation depends only on the characteristics of the tissue, while the T2* signal loss is more rapid and may be dependent on the magnet's physical properties as well (**Figure 4**).

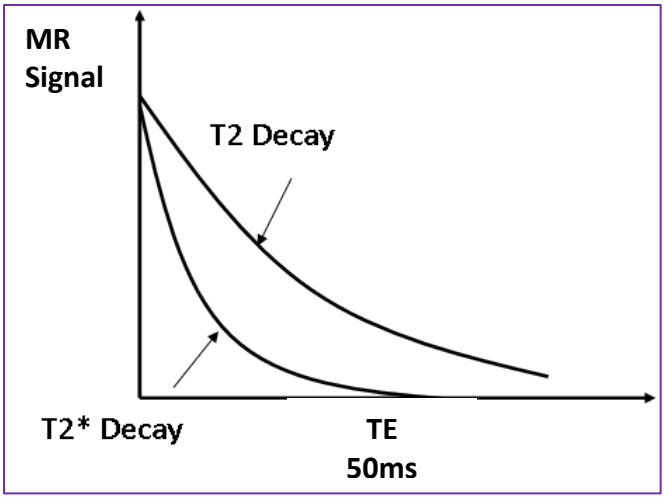

Figure 4. Effect of T2 on rate of signal decay.

The choice of TE determines how much T2 signal loss occurs when the signal is recorded, and this in turn determines the relative brightness of the tissues in the images. When the TE is very short (typically 25ms or less), most tissues have very little T2 relaxation. As a result, almost all tissues will still generate significant signal and brightness on the image that depends primarily on their proton densities. Therefore, the contrast between different tissues in the image will be modest.

When the TE is long (typically 50-120ms), there is time for significant T2 decay. Tissues with short T2 relaxation times will lose most or all of their signal, while tissues with long T2 relaxation times will lose some of their signal but retain much more than short T2 tissues. This is reflected in their brightness on the final image. When the TE is long, tissues with short T2 will appear dark, while tissues with long T2 will appear bright (**Figure 5**).

Longer TE results in more T2 decay, different signal from long T2, and short T2 tissues.

Short TE results in less T2 decay, similar signal from long T2, and short T2 tissues.

Whether bright or dark, tissue appearance is dependent on the T2 relaxation time, and the resultant image is referred to as a **T2-weighted image** (T2W). In general, the longer the TE used to acquire the image, the more T2-weighted it will be.

In a T2-weighted pulse sequence, the TE is chosen so that it clearly differentiates the tissues of interest. If the TE is too short, then all tissues will be relatively bright and indistinguishable. If the TE is too long, signal from even long T2 tissues may be lost and the entire image may be very dark or of poor quality, thus losing the contrast between tissues of interest.

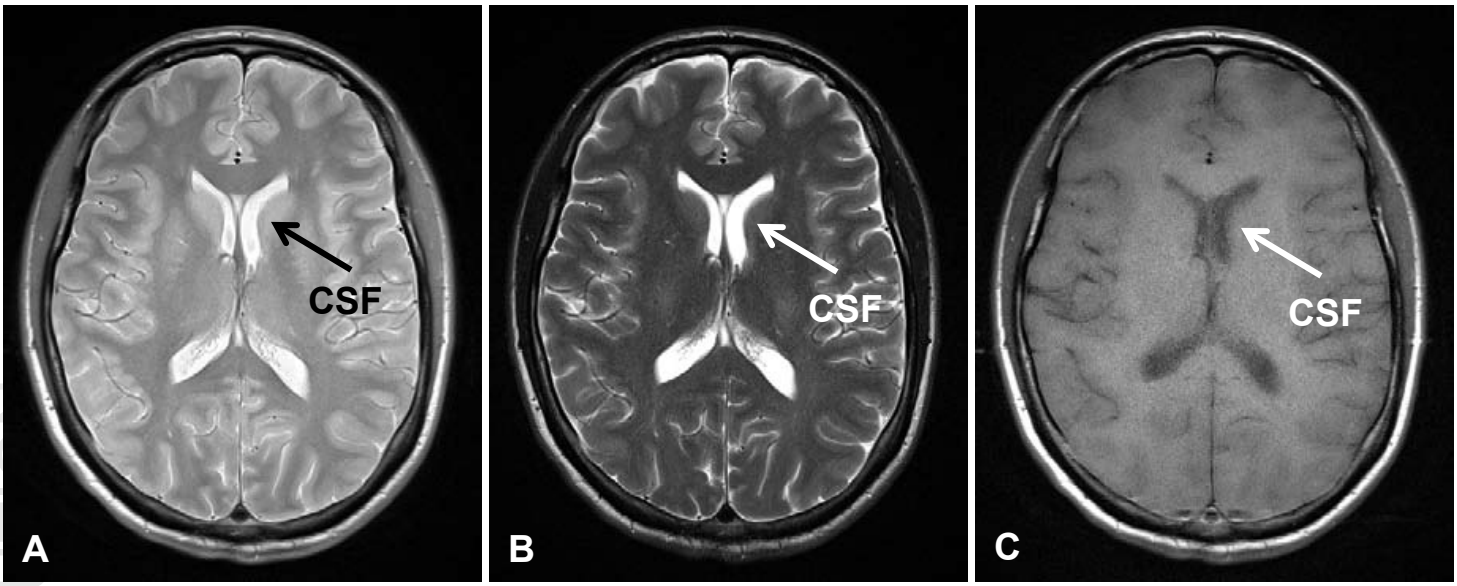

Figure 5. MRI of the brain. (A) Proton density-weighted axial image (TR = 4500ms, TE = 13ms). (B) T2-weighted axial image (TR = 4500ms, TE = 113ms). (C) T1-weighted axial image (TR = 600ms, TE = 13ms).
Courtesy of University of Iowa Carver College of Medicine.

Repetition time

Repetition time (TR) is the second important parameter that determines the relative brightness and contrast of tissues in a spin-echo pulse sequence. TR is the time between each of the 90° pulses in the sequence.

Immediately after a 90° excitation pulse, the longitudinal magnetization goes to zero as all of the magnetization is flipped into transverse magnetization. After the excitation pulse, the process of T1 relaxation begins to restore the longitudinal magnetization. As this relaxation process continues, the longitudinal magnetization continues to grow until the next 90° pulse flips all of the restored magnetization into transverse magnetization. The amount of T1 relaxation that occurs is therefore determined by TR.

Clearly the longer the TR, the more time there is available for T1 recovery of longitudinal magnetization. Since the subsequent 90° pulse flips all of the longitudinal magnetization into transverse magnetization, a longer TR means more magnetization during the next excitation and therefore more signal and a brighter appearance on the image.

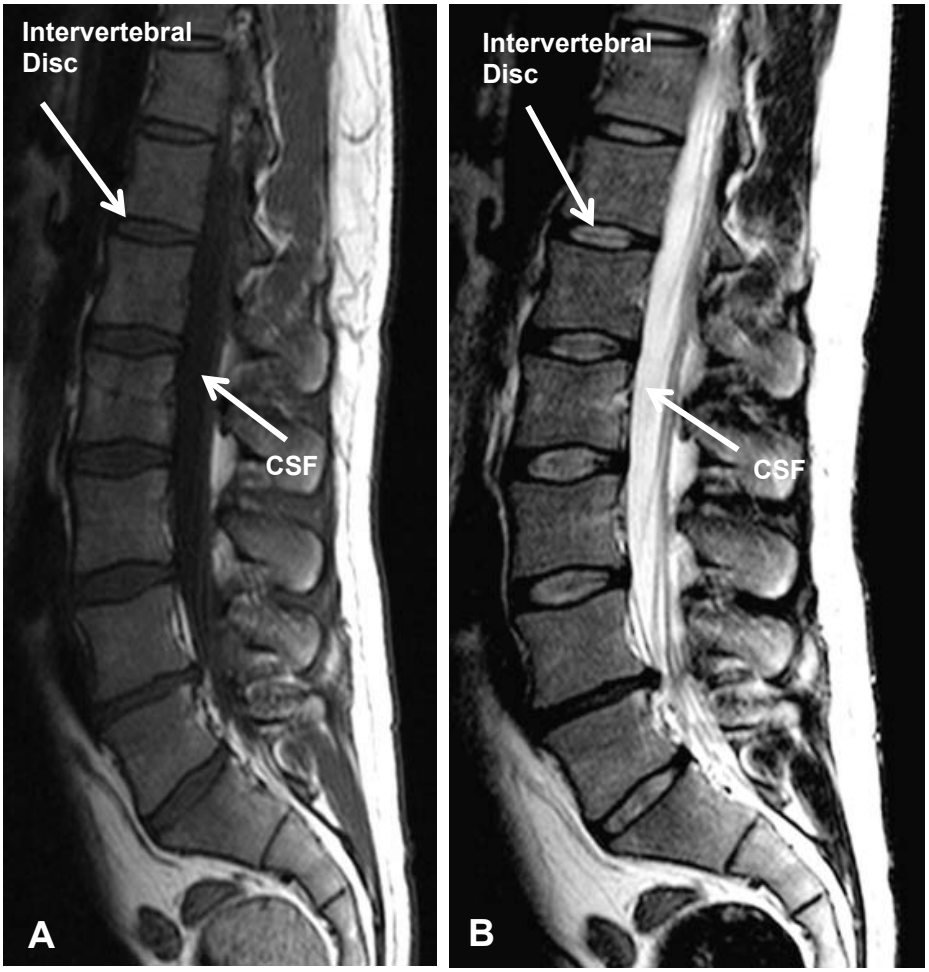

Figure 6. Sagittal images of the spine. (A) T1-weighted. (B) T2-weighted. Note the relative brightness of CSF and intervertebral discs on image B. *Courtesy of University of Iowa Carver College of Medicine.*

If the TR is very long, it allows all tissues — both those with short T1 and long T1 — time to fully recover their longitudinal magnetization before the next 90° excitation. As a result, tissues with both short and long T1 relaxation will appear equally bright on an image. As the TR gets shorter, there is less time for magnetization to recover. While tissues with short T1 recovery times may still have nearly complete restoration of magnetization, those with long T1 recovery will fall short of their full magnetization. As a result, when the next 90° pulse is applied, there will be less signal generated from the tissue with long T1 compared to the tissue with short T1. In the reconstructed image, the tissue with long T1 will be dark, while the tissue with short T1 will be bright. An image using this type of contrast is called **T1-weighted** (**Figures 5-7**).

In a T1-weighted pulse sequence, the TR is selected so that the image will best distinguish between tissues of different T1 relaxation times. If the TR is too short, then none of the tissues will have time to restore magnetization, and the entire image will appear dark and be of poor quality. If the TR is too long, signal from even long T1 tissues will have time to recover, and the whole image will appear bright, with little difference between tissues.

Longer TR results in more T1 recovery, similar signal from long T1, and short T1 tissues.

Shorter TR results in less T1 recovery, different signal from long T1, and short T1 tissues.

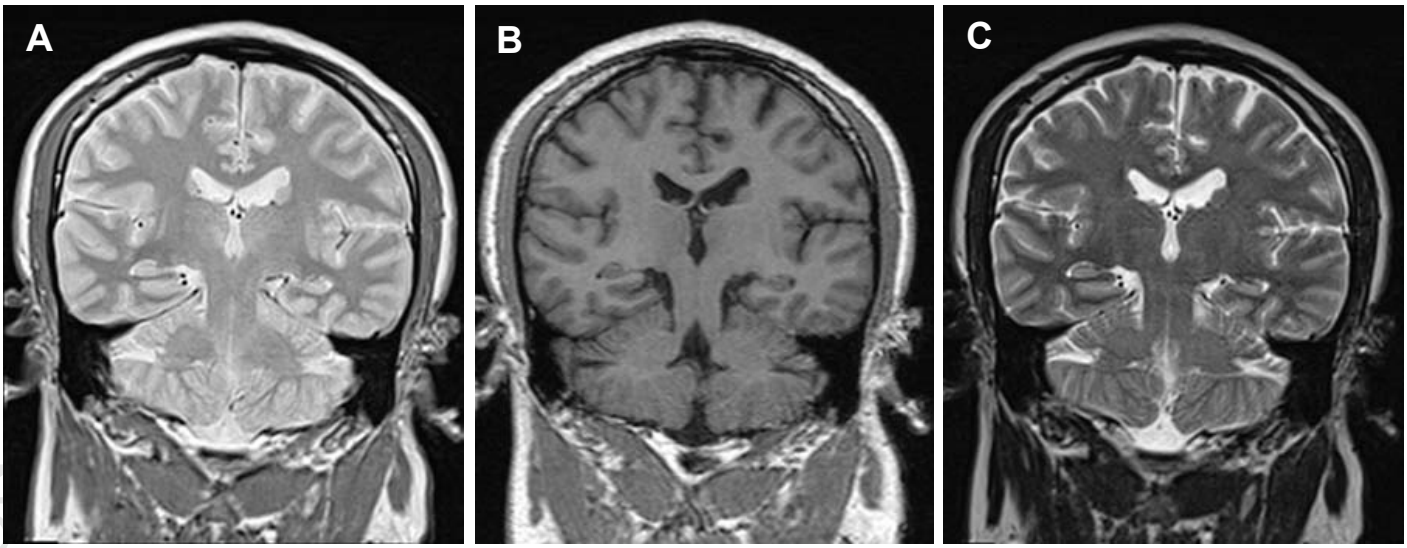

Figure 7. Coronal images of the brain.(A) Proton density-weighted. (B) T1-weighted. (C) T2-weighted. *Courtesy of University of Iowa Carver College of Medicine.*

Typical Parameters for T1, T2, and Proton Density Acquisitions

The parameters TR and TE are both required for spin-echo image acquisition and are selected according to desired type of tissue contrast. For a T2-weighted image, a long TR and long TE are chosen, making the image depend strongly on T2 and very little on T1 relaxation. For a T1-weighted image, a short TR and a short TE are selected, making the image depend mostly on T1 and very little on T2 relaxation.

A combination of short TE and long TR is used in some spin-echo acquisitions, with the image depending very little on either T1 or T2. The short TE does not allow for significant T2 decay, while the long TR permits nearly complete recovery of longitudinal magnetization. In this case, the image is referred to as proton density-weighted (PDW) because the brightness depends mostly on the number of hydrogen protons generating signal at each location (see Figures 5 and 7).

A summary of the combinations of TR and TE is provided in **Table 1.**

	Short TR	Long TR
Short TE	T1-weighted	PD-weighted
Long TE	mixed, not used	T2-weighted

Table 1. Summary of the combinations of TR and TE in spin-echo imaging.

	PD-weighted	T1-weighted	T2-weighted
TR	1500-7500ms	150-750ms	1500-7500ms
TE	5-25ms	5-25ms	50-100ms

Table 2. Typical TR and TE for three types of image contrast in spin-echo imaging.

It should be noted that there are tissues with extremely short T2 and others with extremely long T1 relaxation times. These tissues may appear with reduced signal even on PDW images because the TE or TR cannot be selected to completely eliminate relaxation effects. As a result, all image acquisitions of these tissues will show some T1 or T2 dependence.

Table 2 indicates typical ranges of TR and TE for the three types of image contrast. The ranges are approximate since many factors, such as field strength and RF heating (**specific absorption rate** or SAR), may change the optimum choice for a given exam. Typically, imaging centers select parameters that represent the preferred image quality and contrast for their own equipment and applications.

1.5T VS 3.0T CONSIDERATIONS

With 3.0T scanners becoming more widely available, we need to consider of the effects of higher field strengths on TR and TE. In particular, T1 increases as the field strength increases, while T2 changes very little. Because of the longer T1 relaxation time, simply duplicating imaging parameters used for a 1.5T exam to a 3.0T scanner may result in images that do not have similar contrast. To obtain the same relative contrast as seen at 1.5T, the TR for an exam performed at 3.0T should be longer to compensate for the slower T1 recovery. Parameters may need to be adjusted to compensate for the higher SAR at 3.0T. The change in parameters may affect the scan time or other parts of the imaging protocol. While the same general rules about TR and TE apply, the range of "short" and "long" TR times is slightly different at 3.0T to achieve the same results.

T1 and T2 of Common Tissues

While the actual T1 and T2 relaxation times of many body tissues have been measured, it is not necessary in practice to know the precise relaxation times for specific tissues. Relaxation times may vary due to a variety of factors like magnetic field strength. However, it is useful to have a qualitative understanding of the relative T1 and T2 relaxation of different tissues to predict their appearance on common imaging pulse sequences. As previously noted, tissues with short T1 appear brighter than tissues with long T1; tissues with longer T2 appear brighter than tissues with short T2.

On standard spin-echo images, tissues with short T1 appear brighter than tissues with long T1; tissues with longer T2 appear brighter than tissues with short T2.

MECHANISMS OF T2 RELAXATION

There are a few general principles that predict which tissues will have short or long relaxation times. For example, T2 relaxation results from microscopic variations in the local magnetic field. Signal can be lost because of the dephasing of spins precessing at slightly different rates. In tissues with very mobile hydrogen atoms like pure water, the motion of atoms tends to make these field changes average out, keeping them in phase and resulting in a slow loss of signal, that is, long T2 relaxation.

Hydrogen bound to large immobile molecules cannot move as freely so this averaging effect does not occur. Dephasing and signal loss are also more rapid, that is, short T2 relaxation. **Figure 8A** demonstrates cerebrospinal fluid (CSF) with very bright signal since it has a much higher proportion of free water and therefore a long T2 compared to the surrounding brain tissue.T2 weighting is a valuable tool for showing edema throughout the body.

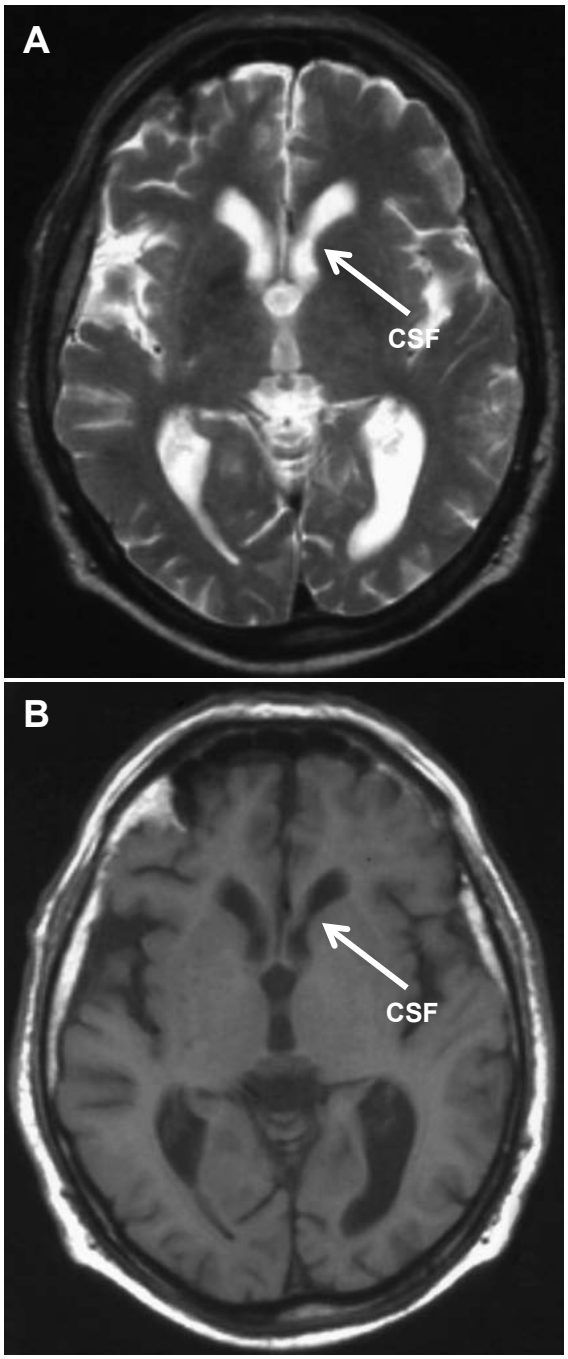

Figure 8. MRI of the brain. (A) Axial T2-weighted image. (B) Axial T1-weighted image at the same level. *Courtesy of Maimonides Medical Center, Brooklyn, NY.*

MECHANISMS OF T1 RELAXATION

T1 relaxation occurs as the excited nuclei give their absorbed energy back to the surrounding lattice, moving spins back to their equilibrium state. The relaxation process results from the local motion of the nuclei. T1 relaxation occurs rapidly when the rate of this motion matches the Larmor frequency. The size of the molecule is one factor that influences this motion. Recall that the Larmor frequency is determined by the main magnetic field, B_0. Small molecules such as water move more rapidly than the Larmor frequency, and large molecules such as proteins move more slowly.

Figure 8B demonstrates some of these T1-weighted characteristics. The subcutaneous layer of fat appears bright while CSF appears dark. Other brain tissues have an intermediate signal because they have a smaller proportion of free water. The dark CSF signal on a T1-weighted image is the opposite of its appearance on a T2 image where it appears bright.

This example demonstrates how multiple images with different T1- or T2-weighting help identify precisely what type of tissue is present. Approximate T1 and T2 relaxation values for several tissues are shown in **Table 3**. As noted previously, actual values may vary; therefore, it is the *relative* value of different tissues that determines how they will appear on T1- and T2-weighted images.

TISSUE	T1 at 0.5T (ms)	T1 at 1.5T (ms)	T1 at 3.0T (ms)	T2 (ms)
Gray Matter	650	950	1350	95
White Matter	550	750	850	85
CSF	2000	3000	4000	500
Muscle	600	1120	1400	45
Fat	200	250	350	80

Table 3. Approximate values for several tissues at three different magnet strengths.

Gradient-echo Pulse Sequence

The other widely used type of pulse sequence in MR image acquisition is the gradient-echo pulse sequence, shown in **Figure 9**.

The gradient-echo sequence looks similar to the spin-echo pulse sequence described earlier. The sequence begins with an excitation pulse; this may be a 90° pulse, or it may use a smaller flip angle. Unlike the spin-echo, the gradient-echo sequence has no refocusing 180° pulse after the initial excitation. After a delay of the duration of the echo time, the signal is recorded and the steps are repeated with a repetition time.

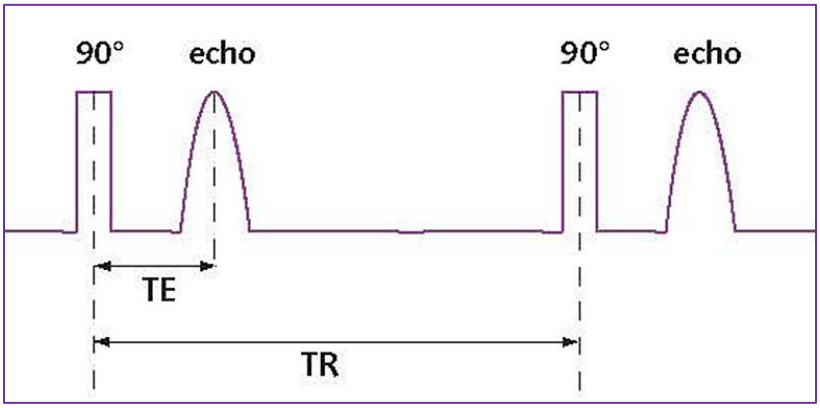

Figure 9. Gradient-echo pulse sequence.

Because the acquisition has no refocusing 180° pulse, there is no compensation for the loss of signal from T2* relaxation. Therefore, the signal recorded depends not on the T2 of the tissue alone but on T2* relaxation (recall that this may be affected by both the tissue and the **inhomogeneities** of the magnetic field).

Effect of TR

As with spin-echo imaging, the TR and TE of a gradient-echo pulse sequence are under the control of the scan operator and can be varied to produce different tissue contrast properties in the MR image. Additionally, the flip angle of the excitation pulse can be varied to adjust tissue contrast.

The elimination of the refocusing pulse results in each repetition being completed more quickly, that is, the TR in a gradient-echo pulse sequence can be much shorter than for spin-echo. Depending on the precise application and protocol, the TR may be anywhere from a few milliseconds to a few hundred milliseconds, and complete gradient-echo images can be acquired at a rate of several images per second.

As with spin-echo, it is still true that a shorter TR results in a more T1-weighted image. The precise appearance depends on the combination of the TR and the flip angle.

Flip Angle

The scan operator can select the flip angle in a gradient-echo sequence. When the TR is long, a 90° pulse is the best choice for obtaining the most signal in the image. However, as the TR gets shorter for faster imaging, a 90° pulse may not be optimal. With a short TR, there is very little time for the longitudinal magnetization to recover from zero for the next excitation, so a 90° excitation results in very low signal.

With a short TR, a lower flip angle leaves some of the magnetization still pointed along the longitudinal axis afterwards. Even after a short TR, there will still be some longitudinal magnetization available for the next excitation. After many repetitions of the excitation, the magnetization will settle into a **steady state** that determines the relative amount of signal that can be achieved. **Figure 10A** illustrates how the flip angle and steady-state signal are related for two combinations of TR and T1.

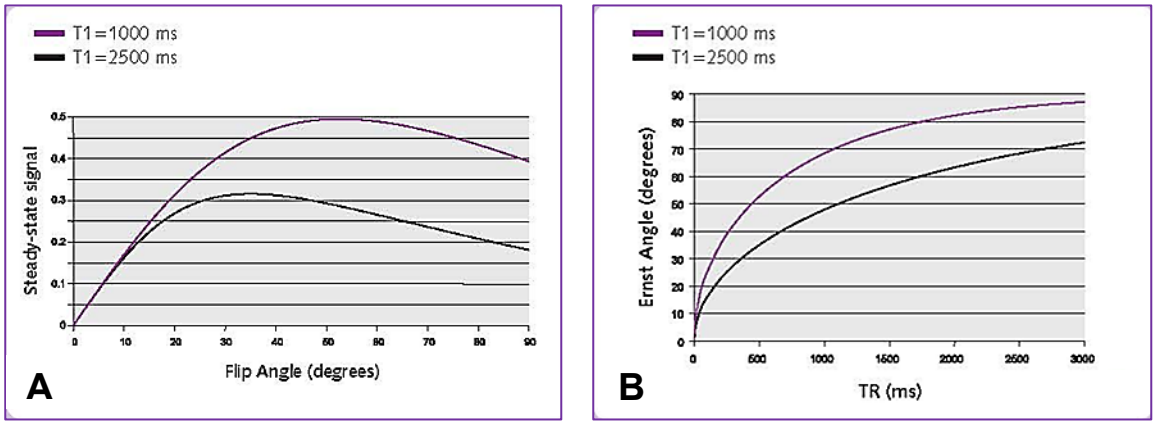

Figure 10. (A) Steady-state signal vs flip angle in gradient-echo for TR=500ms. (B) Ernst angle, the flip angle that gives the maximum amount of signal for a given combination of TR and T1.

Ernst Angle

The Ernst angle is a particular flip angle that generates the maximum signal for a specific combination of TR and T1 and describes the flip angle that generates the largest amount of signal for a particular tissue T1 and pulse sequence TR combination. The use of a flip angle equal to the Ernst angle optimizes the image quality for a particular tissue, especially when performing rapid imaging with very short TR times. **Figure 10B** depicts a plot of the Ernst angle for a given TR for two different tissue T1 times.

Effect of TE

Because the signal decay in gradient-echo imaging is due to T2* rather than to T2, a truly T2-weighted image cannot be acquired. T2* weighting may be system-dependent and is not routinely used for diagnostic purposes. The TE for gradient-echo pulse sequences is almost always very short (a few milliseconds), minimizing the signal loss from T2* and maximizing the signal in the image.

There are a few cases where T2*-weighted imaging is used. For example, an accumulation of iron in an organ, such as the liver, results in a significant shortening of T2* relaxation. Using a gradient-echo pulse sequence with a longer TE will make the image contrast sensitive to this shortened T2*, and the iron-rich areas will appear darker under these conditions. The different state of iron in hemoglobin will also affect T2*, and this effect is used in **functional MRI** (fMRI) for identifying areas of the brain that are active when stimulated. Nevertheless, the TE used in these applications is still relatively short compared to the TE in a T2-weighted spin-echo pulse sequence.

MR Contrast Agents

MRI creates images capable of differentiating among many different tissues based on their T1 and T2 properties. However, it is possible to generate significantly greater contrast in tissue appearance with the use of a **gadolinium-based contrast agent** (GBCA), which has the effect of reducing T1 relaxation time. Tissues that take up the contrast agent will appear brighter on T1-weighted imaging than on noncontrast-enhanced scans because of the greatly reduced T1.

Most MRI contrast agents are based on the rare earth metal gadolinium (chemical symbol Gd). To make it safe for injection, the gadolinium is bound to a complex of other compounds so that it is filtered out of the blood stream within several hours. Gadolinium is **paramagnetic**, with seven unpaired electrons that interact with protons in nearby water molecules to dramatically shorten the T1 relaxation time. The appearance of tissue into which the contrast agent penetrates appears brighter on T1-weighted imaging relative to noncontrast-enhanced tissues. Most commonly, gadolinium-based contrast agents are injected into the bloodstream, and strongly T1-weighted images are acquired. Blood vessels appear much brighter than any other tissue, resulting in an MR **angiogram**, an image that highlights the vasculature.

SAFETY OF GADOLINIUM-BASED CONTRAST AGENTS

The first gadolinium-based MRI contrast agent was approved by the FDA in 1988, and hundreds of millions of contrast doses have been given with minimal side effects since that time.

However, in 2006 the FDA issued an advisory that gadolinium-based contrast agents may be associated with a severe and sometimes fatal condition called **nephrogenic systemic fibrosis** (NSF). A few hundred cases of this condition were confirmed, and all cases were associated with patients who had poor kidney function. As a result, gadolinium-based contrast agents carry a warning about the risk of NSF in patients with compromised kidney function.

Imaging centers should verify that patients who are candidates for GBCA-enhanced MRI exams have adequate kidney function to minimize the risk of developing this rare condition. Professional societies have developed guidelines for screening subjects at risk. When these precautions are taken, GBCAs are safe and effective and are still a vital and widely used tool for contrast-enhanced MRI examinations. **Since the implementation of these guidelines and general increased awareness of the potential for this condition, NSF has been virtually eliminated.**[1,2]

For the most recent information on this potential adverse effect, visit the Food and Drug Administration website at: http://www.fda.gov/Safety/MedWatch/SafetyInformation/SafetyAlertsforHumanMedicalProducts/ucm225375.htm.

In 2012, The American College of Radiology updated its Manual on Contrast Media which prescribes the safe utilization of GBCAs, available at: http://www.acr.org/~/media/ACR/Documents/PDF/QualitySafety/Resources/Contrast%20Manual/2013_Contrast_Media.pdf/#2013_Contrast_Media_Manual.indd:.27738:10181.

Summary

The two fundamental pulse sequences that are the basis for nearly all MR imaging acquisitions are the spin-echo and gradient-echo pulse sequences. The spin-echo sequence combines a 90° excitation and a 180° refocusing pulse to generate images with T1-, T2-, or PD-weighted tissue contrast. The specific tissue contrast of the image is determined by the setting of the TE and TR, which are under the control of the scan operator. The gradient-echo pulse sequence eliminates the refocusing pulse, creating images with T1-, T2*-, or PDW tissue contrast. The TR and TE are variable in this sequence, as is the flip angle. While a T2-weighted image cannot be generated with a gradient-echo, the shorter sequence can be used to acquire images very rapidly for applications such as cardiac imaging.

The contrast in an MR image can also be changed and enhanced by the use of gadolinium-based contrast agents. GBCAs act to shorten T1 of the hydrogen nuclei, resulting in tissues appearing brighter on T1-weighted imaging.

Starting with these basic pulse sequences, there are a variety of changes that can be made to create MR images whose brightness depends on a broad range of factors, such as blood flow, chemical composition, and diffusion of molecules.

POINTS for PRACTICE

1. Why does a 180° excitation pulse follow the 90° excitation pulse in spin-echo?
The 90° pulse flips the magnetization completely into the transverse direction to generate maximum amount of signal. The 180° pulse refocuses the signal loss of T2* by flipping the orientation of the magnetization around so spins that have rotated ahead of the others are now lagging by the same amount, and spins that have fallen behind are now ahead of the rest. The 180° pulse rephases transverse magnetization, leading to the formation of an echo. After the 180° pulse reorients the spins, the dephasing or "fanning out" process begins to reverse itself, with the spins coming back towards each other until they are all oriented in the same direction as they were after the 90° pulse. This generates the maximum possible signal and eliminates T2* effects.

2. Define TE and TR, two important time constants associated with spin-echo pulse sequences.
TE, or echo time, measures the time between the initial 90° pulse and the time at which the signal is actually recorded. TR, repetition time or recovery time, is the time between each of the 90° excitation pulses in the sequence.

3. How do TE and TR influence the appearance of tissues in spin-echo pulse sequences?
Over the duration of TE, both T2 and T2* signal loss occur. Shorter TE produces the maximum signal from the tissues and appears bright; longer TE allows for decay, resulting in greater contrast for tissues with short T2 appearing dark and tissues with long T2 appearing bright. Over the duration of TR, T1 relaxation occurs. If TR is long, all tissues recover their magnetization and appear bright. If TR is short, some tissues do not fully recover their magnetization and appear darker than tissues with short T1. Tissues with long T1 do not fully recover and appear darker than tissues with short T1.

4. What are T1 and T2 weighting?
In T2-weighted images, the brightness of the tissue is mainly determined by its T2 relaxation time. In T1-weighted imaging, it is the T1 relaxation time that determines brightness. As a result, the same tissues may appear different on each type of image.

5. The scan operator may choose parameters for T1, T2 and proton-density weighted images. What is a general rule for making these selections?

- For a T2-weighted image, select a long TR and long TE.
- For a T1-weighted image, select a short TR and short TE.
- For a PD-weighted image, select a combination of long TR and short TE.
- On a T1-weighted image, subcutaneous fat typically has high signal intensity (brighter), and fluids have low (darker) signal intensity.
- On a T2-weighted image, subcutaneous fat has medium signal intensity and fluids have high signal intensity.

6. TR and TE are two important time constants also associated with gradient-echo pulse sequences. Explain why. What is the role of the flip angle in a gradient-echo pulse sequence?
The TR in a gradient-echo pulse sequence can be much shorter than for spin-echo. A shorter TR results in a more T1-weighted image. With a very short TR, there is little time for the longitudinal magnetization to recover, so here a 90° excitation will result in a very low signal. Also, with a short TR, a lower flip angle will leave some of the magnetization still pointed along the longitudinal axis, leaving some available for the next excitation. The TE for gradient-echo pulses sequences is therefore almost always very short. Because the signal decay in gradient-echo imaging is due to T2* rather than T2, a truly T2-weighted image cannot be acquired. T2* weighting may be system-dependent so is not routinely used for diagnostic purposes. A short TE minimizes the signal loss from T2* and maximizes the signal in the image.

7. What particular kind of flip angle does the Ernst angle describe?
The Ernst angle generates the largest amount of signal possible for a particular tissue T1 and pulse sequence TR combination in gradient-echo imaging, optimizing the image quality for a particular tissue, especially when performing rapid imaging with very short TR times.

8. Explain how gadolinium-based contrast agents interact with water molecules.
The gadolinium (Gd) chelate is paramagnetic and has seven unpaired electrons that interact with the protons in nearby water molecules to dramatically shorten T1 relaxation time. The appearance of tissue where the contrast agent penetrates will appear much brighter on T1-weighted imaging than will noncontrast-enhanced tissues.

NOTES

POINTS for PRACTICE

1. **What is a gradient field? What is the difference between a gradient coil and an RF coil? How do they work together?**
2. **How many sets of gradient coils are there in a clinical scanner? What are their orientations?**
3. **How do gradients generate location information for the image?**
4. **How are slice selection and slice thickness determined using gradients?**
5. **What is the difference between a frequency-encoding gradient and a phase-encoding gradient?**
6. **How are 2D and 3D images created?**
7. **Explain *k*-space and Fourier transform.**

IMAGE FORMATION

The signal recorded in an MRI exam is detected by one or more receiver coils. This signal is a summation of the magnetization of the nuclei that have undergone excitation. The receiver coils alone cannot determine the location of all the nuclei generating this signal, so an additional set of magnetic fields known as **gradient fields**, or gradients, is required to make an image or "map" of the hydrogen nuclei in the body.

Gradient Fields

A gradient field is a magnetic field with a strength that changes depending on location within the magnet. Gradient fields are generated by **gradient coils** which are electromagnets arranged in pairs surrounding the inside bore of the magnet (**Figure 11**). The magnetic field generated is in the same direction as the main B_0 field, so the gradient fields are either added or subtracted from the main magnetic field.

Gradient coils create magnetic fields with strength directly proportional to the distance from the center of the magnet. That is, for every centimeter away from the center of the magnet, the magnetic field increases or decreases by a proportional amount.

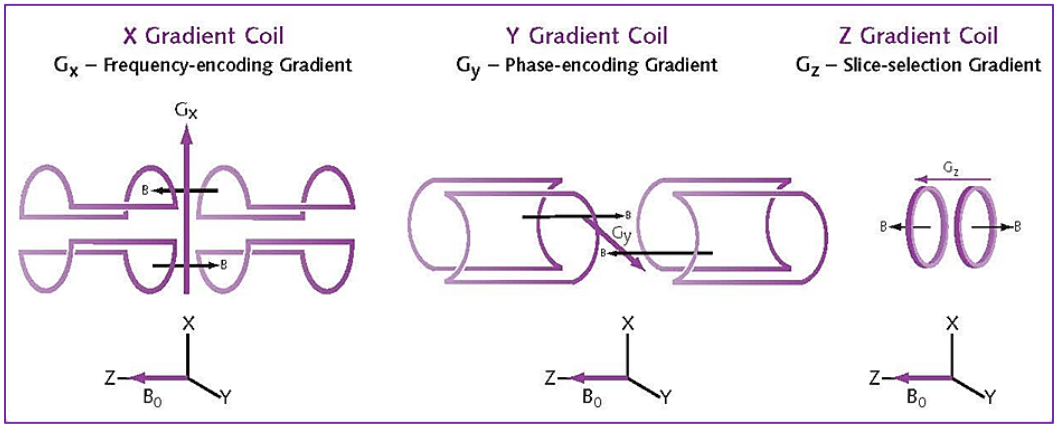

Figure 11. Gradient coils in three orientations. The x, y, and z gradient coils produce magnetic gradients in three perpendicular directions.

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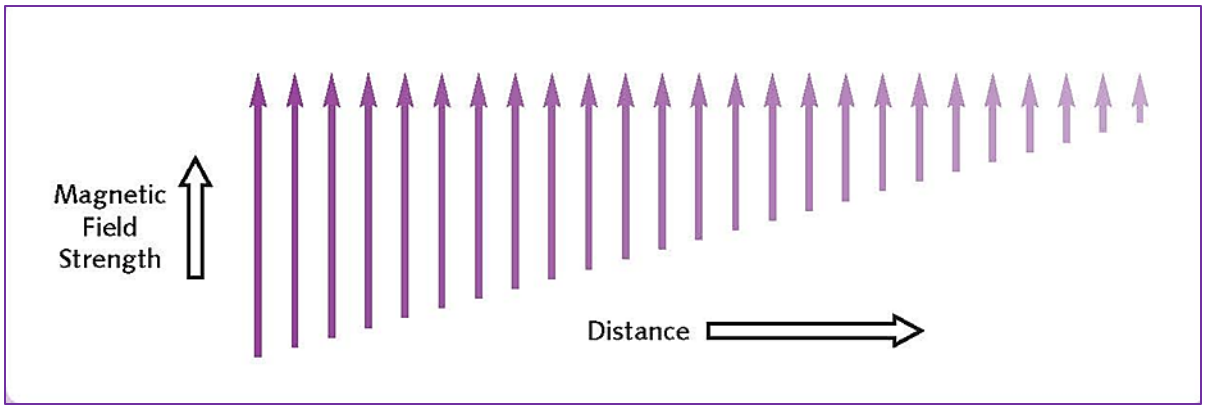

Figure 12. A magnetic field gradient is a magnetic field that steadily increases and decreases in strength with distance.

If the distance from the center doubles, then the change in the magnetic field also doubles (**Figure 12**). The amount of magnetic field change is programmable so that the magnetic field experienced at any location can be controlled.

X, Y, and Z Directions

There are three sets of gradient coils, and each set can be controlled independently. Each coil produces a gradient field oriented along one of the three primary directions: the *x*, *y*, and z axes. As one looks down the bore of the magnet, the *x* direction is left to right, the *y* direction is bottom to top, and the z direction is back to front (**Figure 13**).

The orientation of the gradient sets the direction along which the magnetic field changes as the distance from the center of the magnet increases. For example, the *x* gradient creates different magnetic fields depending on the distance to the left or right of center of the magnet, but the field does not change from bottom to top or front to back. Similarly, the *y* and z gradients only affect their particular directions. However, because the gradients can be operated simultaneously, turning on more than one gradient at a time can make the magnetic field change along a diagonal direction ("single **oblique**" or not 90° to one of the planes) or along any direction ("double oblique" or two angles that are not 90° to the other plane). In all of these cases, the gradients are adding to or subtracting from the main B_0 magnetic field.

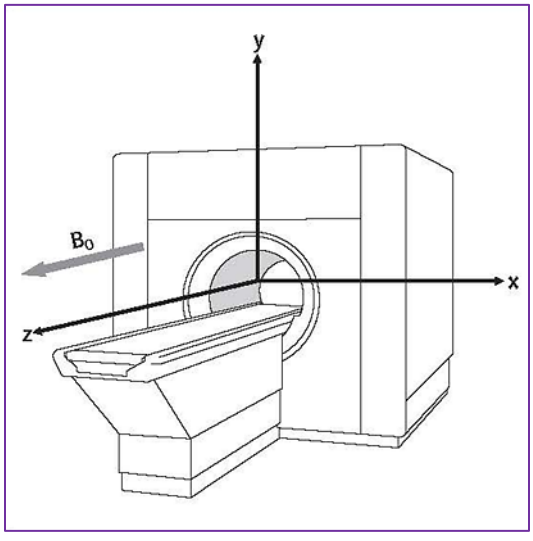

Figure 13. The three primary directions in the closed bore.

The gradient fields make it possible to obtain spatial information from the signals recorded after excitation. The gradients change the magnetic field depending on position within the magnet. Recall that the frequency at which hydrogen protons spin or precess depends on magnetic field strength experienced by the protons. Combining the direct relationship between location and gradient magnetic field with the relationship between magnetic field and frequency means that the gradients will cause the precessional frequency of the spins to directly depend on their location in the magnet. As described below, this change in signal frequency by location allows the scanner to perform slice selection to produce images from thin sections of the body, as well as resolve the location of the bright and dark (high and low MRI signal) areas within the section. It is the rapid switching of the gradient coils that accounts for the "banging" noises heard during an MR exam.

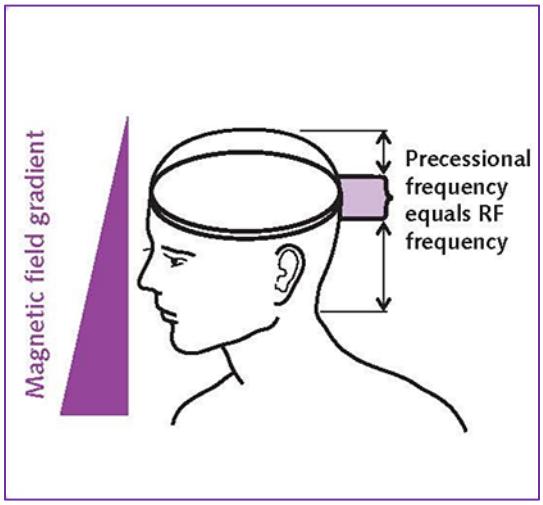

Figure 14. Slice selection is performed by imposing a magnetic field gradient over the patient.

Slice Selection

The purpose of slice selection is to create an excitation that causes only a narrow section or slice of the tissue to experience the full effect of the RF pulse and thus generate signal used to form the image (**Figure 14**). This is achieved by turning on a gradient in the direction *perpendicular* to the desired slice. The effect of this gradient causes the magnetic field strength to vary along this direction. Usually this is shown in the *z* direction, back to front of the scanner bore but in practice might be an arbitrary direction.

Recall that the frequency at which spins rotate, the Larmor frequency, is directly related to the magnetic field experienced as $f = \gamma B_0$. From this equation, it is clear that when a gradient is turned on, the total B_0 changes along the gradient direction, and the Larmor frequency also varies along the same direction as the gradient.

We noted that hydrogen nuclei are excited only if an RF pulse is transmitted at the Larmor frequency. Combining the RF pulse bandwidth, the **slice-selection gradients**, and the desired slice thickness means that transmitting an RF pulse with a specific range of frequencies results in an excitation of a band or slice of tissue at a specific location.

The gradient field determines the actual section (spatial location) that is sensitive to a range of frequencies, that is, that have the Larmor frequency within the same range of frequencies as the excitation pulse.

Bandwidth

The RF pulses transmitted are not usually made up of a single frequency but a combination of a range of frequencies. This range is called the **bandwidth** of the pulse and determines the range of frequencies the pulse contains as well as the frequencies of the hydrogen nuclei that will be excited.

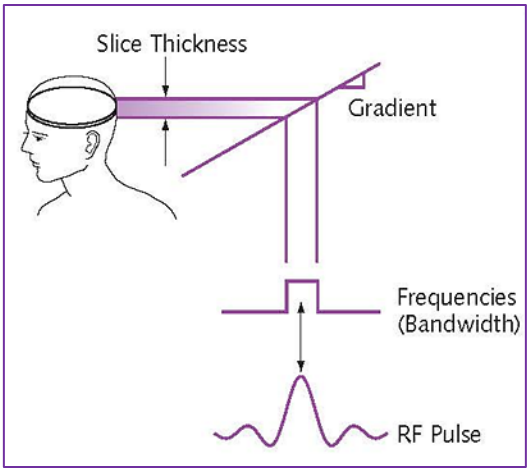

Figure 15. Relationships among bandwidth, gradients, and slice thickness.

The final slice-selection pulse combines an RF excitation pulse with a given bandwidth and a slice-selection gradient to alter the Larmor frequencies based on location in the body. Only the hydrogen nuclei precessing at the Larmor frequency within the bandwidth of the RF excitation pulse are excited, thus generating signal in the final image.

Both the bandwidth of the RF pulse and the strength of the slice-selection gradient are components controlled by the scan operator. When the slice thickness is chosen, the scanner adjusts the RF pulse and gradient parameters to ensure that the correct slice thickness and location are created. A larger bandwidth of the RF pulse results in a thicker slice because a wider range of frequencies is excited. A stronger gradient results in a thinner slice since the range of locations that fall into the bandwidth of the RF pulse is smaller. **Figure 15** illustrates the relationships among bandwidth, gradients, and the excited slice.

Spatial Location

Frequency Encoding

The use of a slice-selection pulse yields only one direction of spatial information by selecting a single plane oriented perpendicular to the gradient used for slice selection. In order to create a two-dimensional image or possibly a three-dimensional volume, additional spatial information is required. The frequency-encoding gradient is applied in an orientation different from the slice selection so that the spatial information gained is along a perpendicular direction. By convention, this is usually shown in the *x* direction in diagrams of the slice, though it could be in any direction.

Once the slice-selection gradient is turned off, all of the excited spins in the slice spin at the same rate, again experiencing the same magnetic field B_0. During the recording of the signal from the excited slice, a different gradient orientation can be turned on, just as for slice selection. This gradient yields spatial information along a perpendicular direction. Turning on this gradient changes the field experienced by the nuclei and the Larmor frequency of the spins based on their location because the relationship $f = \gamma B_0$ still applies.

Frequency-encoding Gradient

The signal received from the slice by the receiver coil is the sum of the signal from all of the nuclei. With the gradient field turned on, this signal becomes a combination of signals at many different frequencies, with the frequency of each nucleus depending on its location along the direction of the gradient.

This gradient is referred to as a frequency-encoding gradient since the spinning frequency of the nuclei correlates with their location within the body. The overall process of turning on a gradient and simultaneously recording the signal given is called **frequency encoding** and provides one dimension of spatial information.

Fourier Transform

After the signal is recorded, it is processed by the image reconstruction computer; the relative number of nuclei that are spinning at different frequencies can be extracted using the signal's mathematical properties. Because of the direct relationship between the frequency and location of nuclei, the nuclei spinning at a particular frequency can be mapped back to a point along the gradient where they must be located (**Figure 16**). The mathematical operation that extracts the frequencies contained in a signal is called a **Fourier transform** (FT).

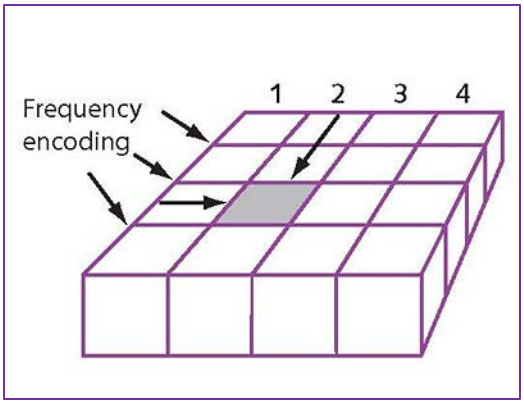

Figure 16. Schematic tissue slice of 16 volume elements (voxels). Frequency encoding determines location along one of the directions.

Phase Encoding

In addition to frequency encoding, another dimension of spatial information is needed to create a two-dimensional image of a single slice of the body. This **phase-encoding** gradient orientation is perpendicular to the slice-selection and frequency-encoding direction. By convention, it is usually shown as the *y* direction in diagrams of the slice.

Phase-encoding Gradient

The phase-encoding gradient alters the relative position or phase of the hydrogen nuclei as they spin. Frequency and phase are related but have different properties. Frequency describes the *rate* at which the nuclei rotate. Phase is the *total number* of rotations, or fraction of a rotation, that are made while the nuclei spin.

An auto racing analogy applies: the frequency is the speed at which the cars travel or how many laps per unit of time; the phase is the number of laps or partial laps completed. Also, only the fractional part of a rotation can be determined from each measurement, in the same way that the number of laps that a car has completed cannot be determined from a single snapshot.

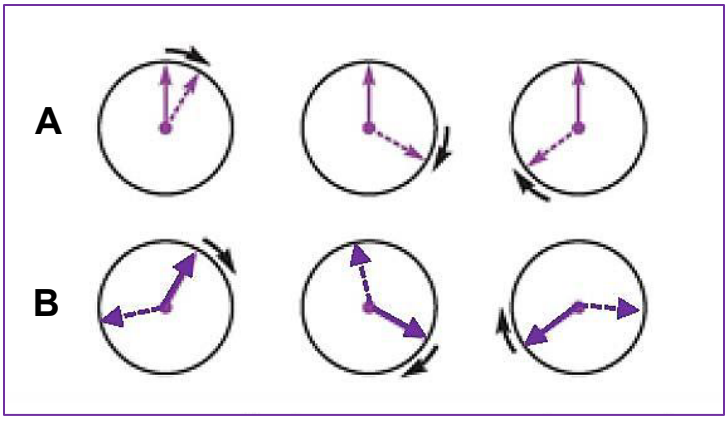

Figure 17. (A) After RF excitation, hydrogen nuclei precess in phase (solid arrows). The phase-encoding gradient causes them to precess at different frequencies and move out of phase (dotted arrows). (B) After the phase-encoding gradient is turned off, hydrogen nuclei again precess at the same frequency but remain out of phase.

The phase-encoding gradient is turned on just after excitation and just before the frequency-encoding gradient and signal readout. While the gradient is turned on, the nuclei spin faster or slower (spin at a higher or lower frequency) depending on their location. As a result, slower spins fall behind, and faster spins move ahead (**Figure 17**).

How far ahead or behind the spins become depends on the magnetic field they experience during the time the phase-encode gradient is turned on. The field is related to location, meaning the relative rotation or phase also depends on location of the spins.

The imaging process continues with the frequency-encoding gradient, but because the extra gradient is turned on for phase encoding, the spins along this phase-encoding direction have been “marked” by the amount of phase change experienced.

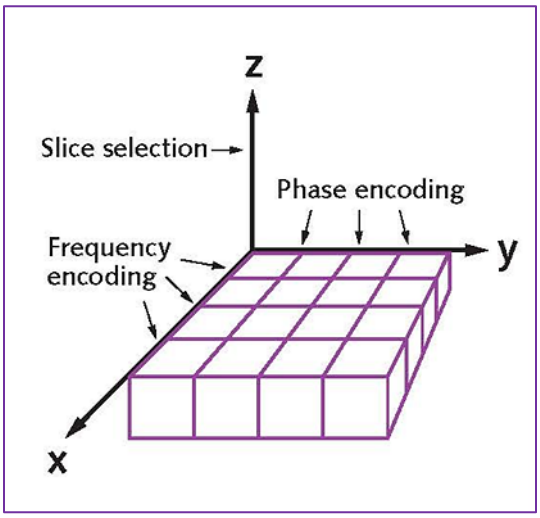

Figure 18. Phase and frequency encoding combine to create a two-dimensional image. The z axis is used for slice selection.

To obtain sufficient information to determine the complete map of hydrogen nuclei along this third phase-encoding direction, the steps are repeated with different strengths of the phase-encoding gradient. Each phase-encoding step acquired by repeating excitation and frequency encoding with different phase-encoding gradient strengths provides one more line of information in the final image. The number of different phase-encoding steps used usually determines the number of points along that direction in the final image. For example, an image with 256 lines typically requires 256 phase-encoding steps. The orientation of the three gradient directions is shown in **Figure 18**.

For two-dimensional imaging, a single image for the excited slice is produced. However, it is also possible to create multiple images or a 3D volume from the selected slice, achieved by using frequency encoding for one direction as before and phase encoding for two directions.

To obtain all of the required spatial information for a 3D volume, all combinations of phase-encoding steps in each of the two directions are needed. For example, to create 256 lines of information along the *y* phase-encoding direction and 64 lines of information along the *z* phase-encoding direction requires 256 × 64 or 16,384 steps.

The Complete Imaging Process

We have just explained all of the steps needed to create two- or three-dimensional images. **Figure 19A** illustrates the order and timing of these steps on a single pulse-sequence diagram. The first step is to perform slice selection with an RF excitation pulse and a *z* gradient to excite nuclei in a narrow slice of the body. Next, the phase-encoding gradient along the *y* axis is turned on for a fixed duration to create information about the spatial location of nuclei in this direction. Finally, the *x* or frequency-encoding gradient is turned on while the signal is recorded for one line of the image. This is illustrated in **Figure 19** as data acquisition (DAQ). The complete image is acquired by repeating this process for all required phase-encode lines as in **Figure 19B**.

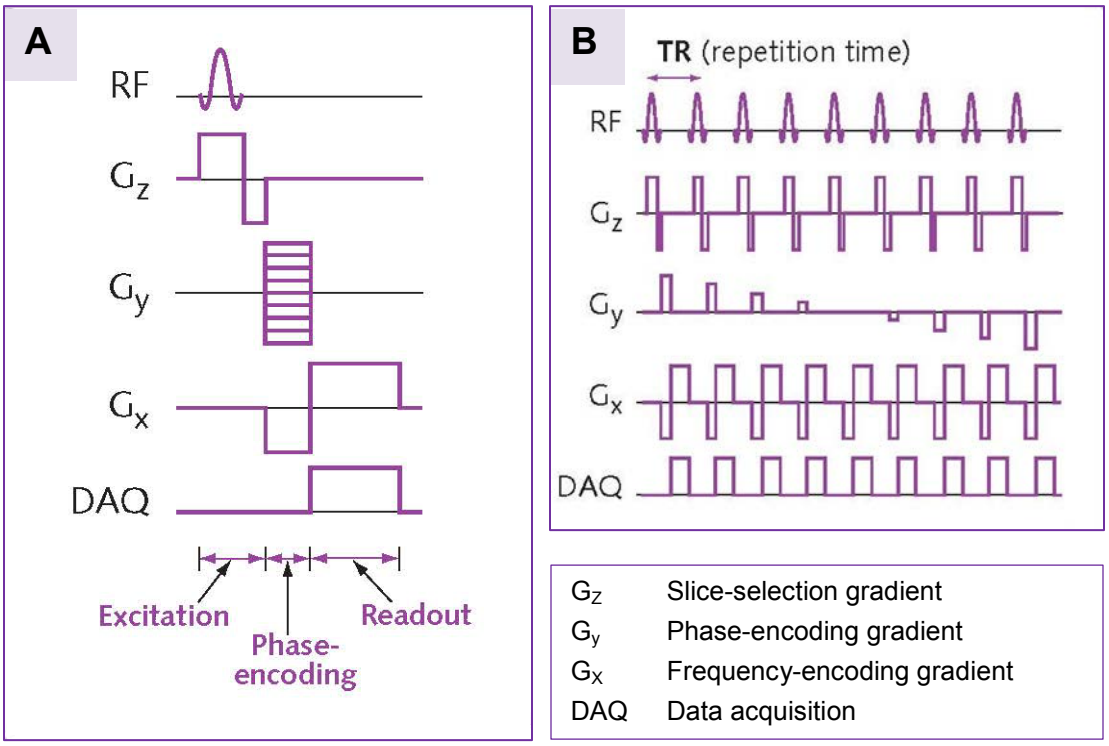

Figure 19. (A) Typical pulse-sequence. (B) Acquisition of a complete image.

While our illustrations use *x* as the frequency-encoding direction, *y* for the phase-encoding direction, and z for slice selection, the directions of each of these gradients can be selected in any oblique orientation by turning on the appropriate combination of gradients at the same time. As a result, image orientation can be appropriately selected to best demonstrate the anatomic regions of interest.

k-space

While we described image acquisition in terms of "lines" of the image, the actual data recorded as it comes from the receiver coils look nothing like the final image of the anatomy. An image will typically record from 128-512 points during frequency encoding and signal readout and repeats the process for 64-512 phase-encoding steps. This generates a grid or matrix of data points with a specified number of points in each of the frequency- and phase-encoding directions. Each dimension of the matrix can be set independently.

The signals received are not a direct map of distribution of the nuclei. They are acquired in a data set called ***k*-space**, which represents the summation of signal from all of the nuclei.

Additional processing is required to convert the summated signal into a useable image. Recall that the Fourier transform is the mathematical operation that accomplishes this conversion, and complex computer processing techniques are required.

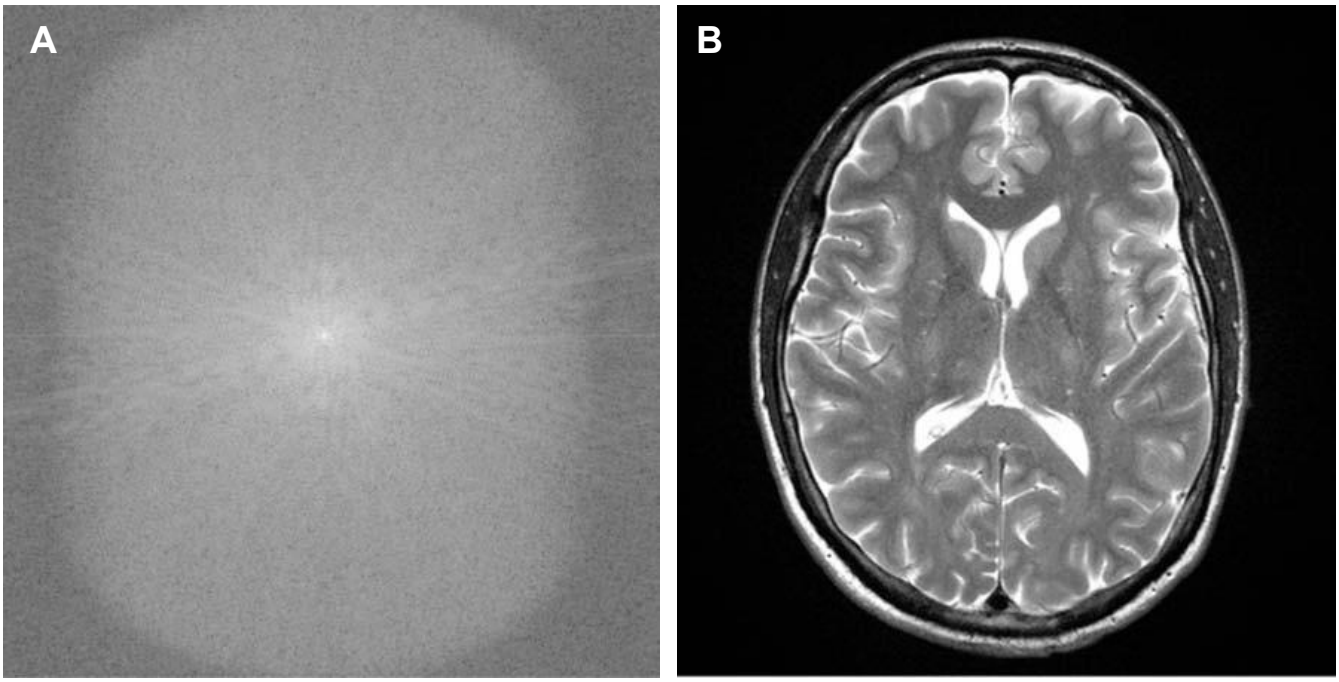

Figure 20. (A) *k*-space representation. (B) A complete image. *Courtesy of University of Iowa Carver College of Medicine.*

The actual *k*-space or "raw" data acquired during an MRI scan as compared to a complete image are shown in **Figure 20**. The signal intensities at different locations relate to the variable distribution of the nuclei over the slice imaged.

The areas near the center of *k*-space represent the low-frequency components of the data where the overall contrast of the image is determined. The areas at the outer margins of *k*-space represent the high-frequency components and provide information on the fine details of the image. The amount of data acquired in these regions determines both the resolution of the image and the sharpness of details. **Figure 21** shows the images that would be generated using only the central or outer areas of *k*-space. The central areas of *k*-space reveal a blurry version of the original image, while the outermost areas of *k*-space have poor contrast but define edges and detail in the original image.

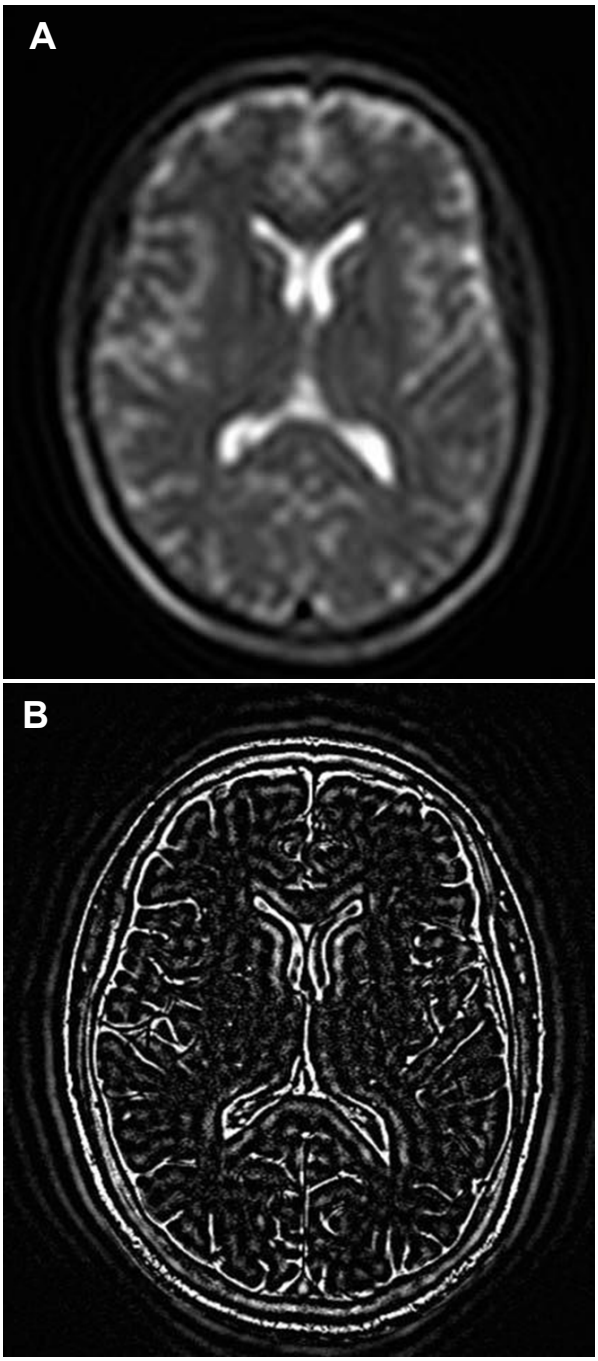

Figure 21. (A) Axial brain image using only central areas of *k*-space. (B) Image using only outermost areas of *k*-space. *Courtesy of University of Iowa Carver College of Medicine.*

We have now described how 2D and 3D images are generated from the MR signal. Gathering spatial information requires a set of gradient magnetic fields that causes the field and frequencies to vary depending on the location of nuclei.

Gradients are used in all aspects of spatial localization. A slice-select gradient is used to excite a thin section of the body. This is followed by a phase-encoding gradient that creates a change in the signal along one direction for a 2D image or a combination of directions for a 3D image. Finally, the frequency-encoding gradient creates the changes needed to locate distributions of nuclei along the final dimension. The process is repeated as many times as needed to acquire the desired matrix size and resolution. The final image is produced after processing the data using the Fourier transform.

These steps are common to virtually all MRI acquisitions. However, with some small changes to the pulses used or to their timing, a wide range of contrast in these images can be created with MRI.

POINTS for PRACTICE

1. What is a gradient field? What is the difference between a gradient coil and an RF coil? How do they work together?
A gradient field, or gradient, is a magnetic field with a strength that changes depending on the location within the magnet. Magnetic fields are generated by gradient coils, which are electromagnets arranged in pairs around the inside bore of the magnet. This type of coil differs from an RF coil, which can transmit the energy for excitation and detect the signal recorded in an MR image. Because RF coils cannot determine the location of all the nuclei that are generating signal, gradients are required to "map" the hydrogen nuclei in the body.

2. How many sets of gradient coils are there in a clinical scanner? What are their orientations?
There are three sets of gradient coils that are perpendicular to each other. One changes the field left to right, one bottom to top, and one back to front.

3. How do gradients "shape" the image?
Gradients make it possible to obtain spatial information from the signal recorded after the excitation. They can be operated independently or in combination, making the magnetic field change depending on their location within the magnet and making it possible to acquire images in any location and orientation.

4. How are slice selection and slice thickness determined?
The purpose of slice selection is to make an excitation that causes only a narrow section or "slice" of tissue experience the full effect of the RF pulse, generating signal used to form the image and achieved by turning on a gradient in the direction perpendicular to the desired slice direction. The final slice-selection pulse combines an RF excitation pulse with a given bandwidth (a range of frequencies that determine the slice thickness) and a slice-selection gradient to alter the frequency of precession (the Larmor frequency) based on location in the body. Only the hydrogen nuclei spinning at the Larmor frequency contained in the bandwidth of the RF excitation pulse will be excited and thus generate signal in the final image.

5. What is the difference between a frequency-encoding gradient and a phase-encoding gradient?
A frequency-encoding gradient makes the frequency of the nuclei directly depend on their location within the body. The signal received from the slice by the receiver coil is still the sum total of the signal from all of the nuclei added together. With the gradient field turned on, this signal will be a combination of signals at many different frequencies, with the frequency of each nucleus depending on its location along the gradient direction. Frequency encoding provides one dimension of spatial information. A phase-encoding gradient is perpendicular to the slice-selection and frequency-encoding directions and is applied prior to the frequency-encoding gradient. It alters the relative position or phase of the hydrogen nuclei as they spin. While frequency describes the rate at which the nuclei rotate, the phase is the total number of complete or partial rotations made while the nuclei spin.

6. How are 2D and 3D images created?
For two-dimensional imaging, a single image for the excited slice is produced with frequency encoding for one direction and phase encoding for the other. For three-dimensional imaging, one must use frequency encoding for one direction and phase encoding for two directions.

7. Explain *k*-space and Fourier transform.
Image acquisition is often described in terms of "lines," but the actual data from the receiver coils look nothing like the final image we see. *k*-space data represent the summation of signal from all of the nuclei simultaneously. The Fourier transform mathematical operation converts these signals into a recognizable image.

POINTS for PRACTICE

1. **One characteristic of fat protons is that their resonant frequency is slightly lower than the Larmor frequency of water protons. What is this difference called?**
2. **How does inversion recovery change the appearance of some tissues?**
3. **Define FLAIR and STIR.**
4. **Which imaging acquisition methods are widely used to display blood flow?**
5. **Name a more efficient version of spin-echo and explain why it is more efficient.**
6. **Which is the fastest of the gradient-echo-based pulse sequences?**
7. **What type of imaging uses a "fanning" type sequence?**

IMAGING CONTRAST TECHNIQUES

We have discussed the basic MRI pulse sequences : the spin-echo pulse sequence applied a 90° pulse for excitation, followed by a 180° refocusing pulse to generate T1-weighted, T2-weighted, or proton-density weighted images. The gradient-echo pulse sequence has a single excitation pulse with no refocusing pulse, and the excitation may be a 90° pulse or any other flip angle. The spin-echo pulse sequence produces the greatest amount of signal; the gradient-echo pulse sequence may be faster.

Additional MR imaging techniques are widely used for many types of imaging studies. By making small changes to the set of pulses used in image acquisition, the appearance of tissues on MRI can be altered in many ways. By introducing additional pulses or by controlling the timing of pulses, some tissues can be made to disappear completely from the image. With other techniques, only moving tissues such as flowing blood are shown. Compared to other imaging modalities, the image contrast in MRI can be tailored to show what is required to answer a specific diagnostic question without additional hardware or special equipment.

Image Contrast Mechanisms

We have seen examples of images in which tissue brightness is dictated by T1 and T2 relaxation. However, other tissue characteristics can change the appearance of tissues in an image.

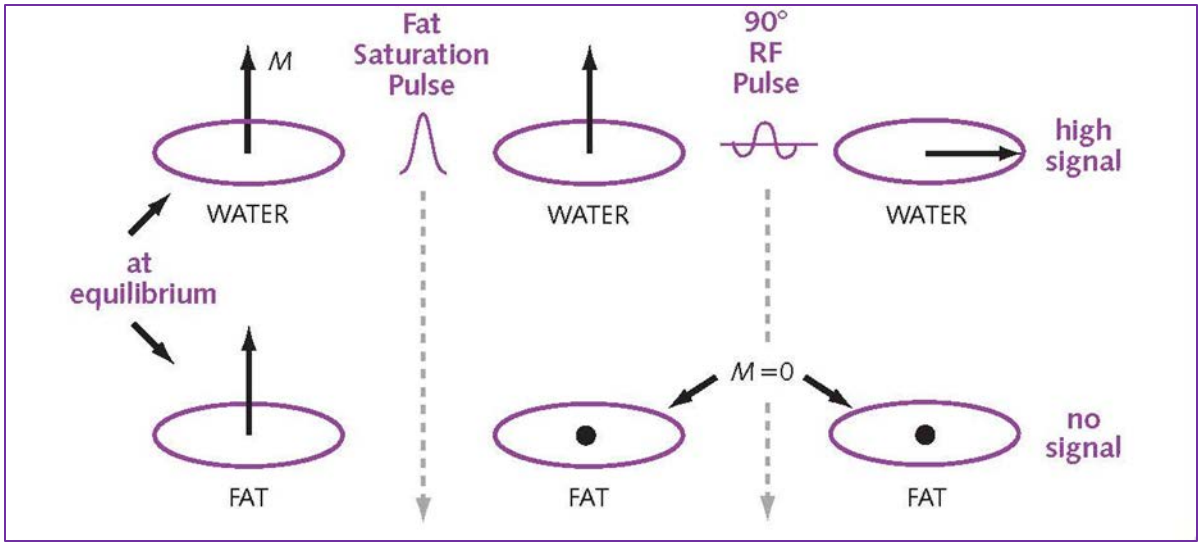

Figure 22. Fat saturation pulse eliminates the magnetization from fat prior to imaging excitation.

Fat Saturation

On conventional spin-echo T1 and T2 images, fat appears bright because of its relatively short T1 and long T2, sometimes making other important tissues difficult to see. One characteristic of fat is that its resonant frequency is slightly lower than that of the Larmor frequency. This difference in resonant frequency is known as **chemical shift** because the chemical properties of fat cause a shift in the frequency at which it resonates compared to surrounding tissues. Because of this difference in frequency, it is possible to configure an excitation pulse specific to the resonant frequency of fat but not of other tissues.

When a 90° pulse is applied to excite only the hydrogen nuclei within fat, all of the longitudinal magnetization of the fat is lost, and all other tissues remain unchanged. This is called a **fat saturation pulse (Figure 22**). An ordinary imaging pulse sequence performed immediately after the fat saturation pulse converts longitudinal magnetization to signal and brightness in the image. For tissues not affected by the fat saturation pulse, their appearance on the image remains unchanged. However, since the longitudinal magnetization of fat is exhausted by the fat saturation pulse, there will be no signal from fat, and it will appear dark on the image**.** Fat saturation is the most common of all fat suppression techniques.

Figure 23 depicts both proton-density and fat-saturated images of a knee. Because bone marrow contains mostly fat, its appearance is bright on the proton-density image but dark on the fat-saturated image. The elimination of the fat signal permits other features of the bone to be seen, like the edema from a bone bruise.

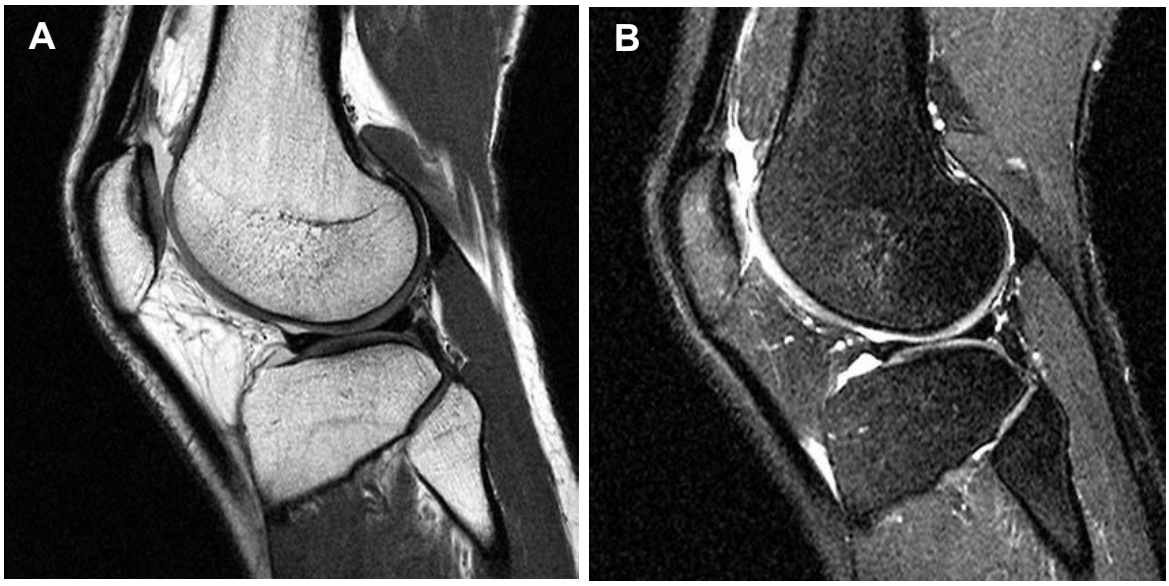

Figure 23. (A) Sagittal proton density-weighted image of a knee. (B) Sagittal T2W fat-saturated image of the knee. *Courtesy of University of Iowa Carver College of Medicine.*

Inversion Recovery

We have also described how to create images with signal that depends on T1 relaxation times by selecting the TR timing. Tissues with short T1 appear bright as compared to tissues with long T1. In other instances, it is desirable to create the opposite effect and make a particular T1 tissue dark or "disappear" completely.

Inversion recovery (IR) technique is used to achieve this effect. Similar to the fat saturation method, the inversion recovery technique works by adding an additional excitation pulse prior to the conventional spin-echo or gradient-echo pulse sequence. In inversion recovery, there is a 180° or inversion pulse prior to the 90° excitation pulse, which flips the magnetization of the nuclei to point in the opposite direction (**Figure 24**). The magnetization remains only along the longitudinal direction so no MR signal is produced by the inversion pulse. Following the inversion pulse, the spins start to recover towards their equilibrium state at a rate dependent upon their T1 relaxation time.

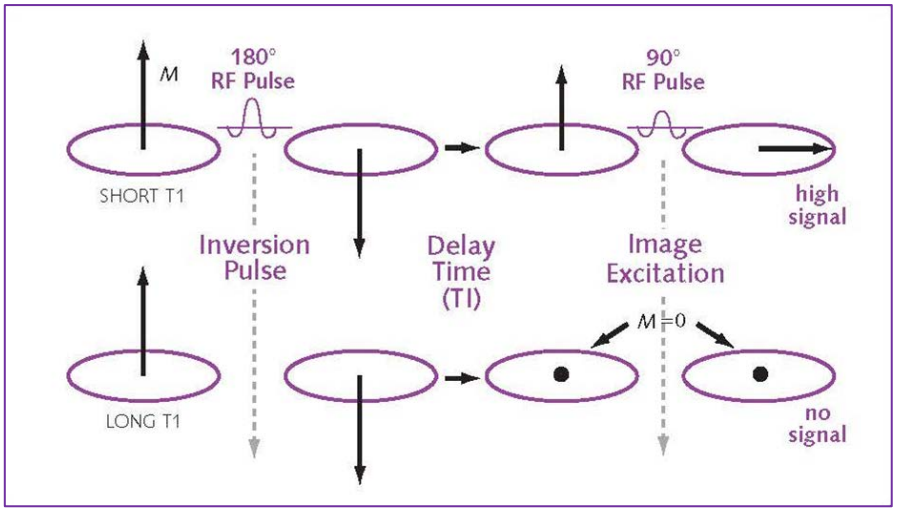

Figure 24. Inversion pulse causes the signal of particular tissues to vanish at the delay time TI (inversion time) immediately prior to image excitation.

At a specific time, the magnetization will be halfway recovered, meaning that the magnetization is zero at that moment (the precise time depends on the T1 relaxation time of the tissue; 70% of the T1 relaxation time is a good approximation). If the imaging pulse sequence steps of excitation and signal readout are performed at this time, then the tissue with zero magnetization will have zero signal and appear completely black in the final image.

The most common uses of inversion recovery are to suppress the signal from fluid, which has a long T1, or fat, which has a short T1.

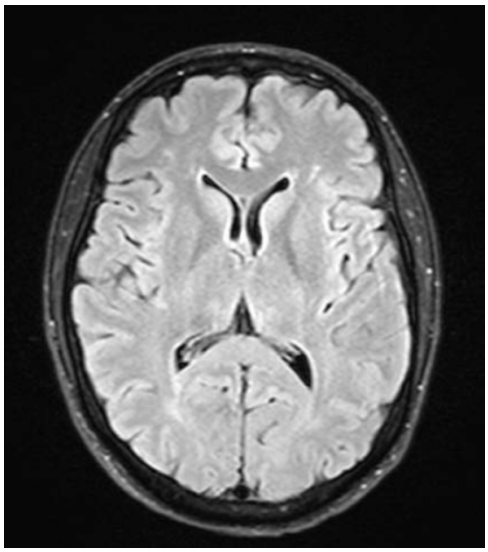

Figure 25. Axial brain image using FLAIR. *Courtesy of University of Iowa Carver College of Medicine.*

FLUID-ATTENUATED INVERSION RECOVERY

A routine application of inversion recovery is to eliminate the signal from cerebrospinal fluid in T2-weighted images of the brain. CSF has a long T1 and T2 compared to other tissues so has a very bright appearance on T2-weighted images, but its magnetization recovers relatively slowly. By using an inversion pulse and waiting until the halfway point in the relaxation of CSF, other tissues will have recovered their signal. The image taken at this point shows most tissue with normal brightness, but the CSF signal disappears and is completely invisible. This use of inversion recovery is referred to as **fluid-attenuated inversion recovery** or FLAIR (**Figure 25**).

Short-T1 Inversion Recovery

Another use of inversion recovery is as an alternative way to make fat signal disappear. The T1 relaxation time of fat is shorter than for most other tissues, so the time at which the fat signal goes to zero is much shorter as compared to fluids. The selected inversion time (TI) is also much shorter and when used this way, the pulse sequence is called **short-T1 inversion recovery** (also short-tau inversion recovery) or STIR (**Figure 26**). STIR may be chosen for fat suppression when imaging with a very large **field of view** (FOV) or where there is metal present in the body as fat saturation may be unreliable in these cases.

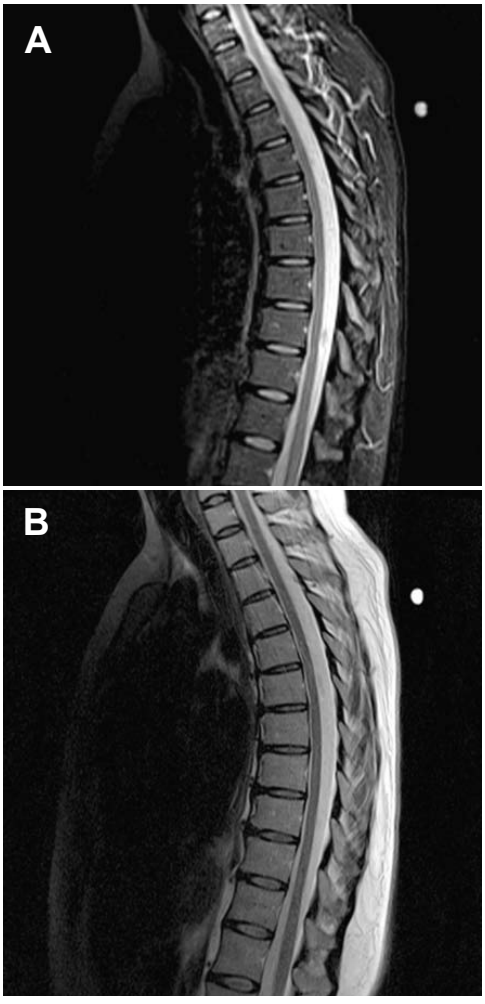

Figure 26. MRI of the spine. (A) Sagittal STIR. (B) Sagittal T2W. *Courtesy of University of Iowa Carver College of Medicine.*

Flow

Time-of-Flight

The most widely used method for viewing flowing blood on MRI is **time-of-flight** imaging (TOF). TOF imaging makes stationary materials appear dark on the image, while moving tissues like flowing blood appear bright (**Figure 27**).

This is accomplished by using pulse sequences that repeat excitation pulses very rapidly so that very little T1 recovery occurs. In stationary tissues, this causes the magnetization to be continually "knocked down" or saturated, as in the fat saturation technique. As a result, these stationary tissues have very little signal in the image and will appear dark. However, since flowing blood is continuously replaced in the imaged area, the inflowing blood will not yet have experienced the saturating excitation pulses and will still generate larger amounts of signal and appear bright.

The resulting image reveals only flowing blood, which in turn provides an outline of the blood vessels. Most often, these images are reconstructed into an angiogram, an image of blood vessels created by combining the many individual sections into a single **maximum intensity projection** (MIP) viewed at different angles (**Figure 27**). In some cases, an MRI contrast agent is used to further highlight the difference between the signal of blood and surrounding tissues (**Figure 28**).

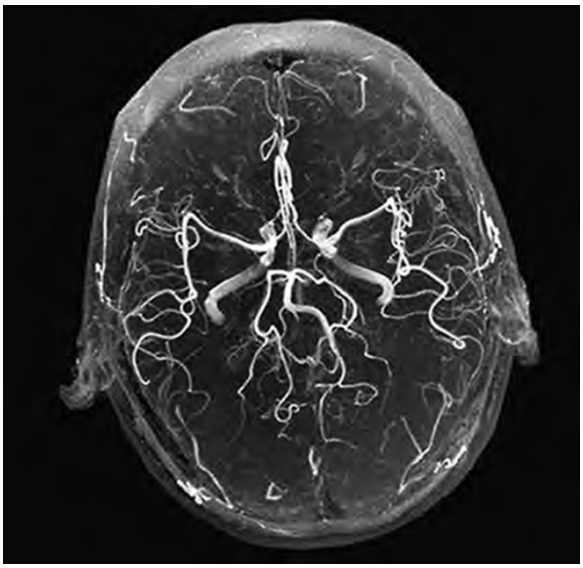

Figure 27. Collapsed maximum intensity projection (MIP) of the intracranial arteries. This time-of-flight (TOF) image takes advantage of the flow-related enhancement of the fast-flowing blood. The individual source images are displayed as a singular MIP projection to visualize the entire vasculature.
Courtesy of Thomas Schrack, BS, ARMRIT, Fairfax Radiological Consultants, Fairfax, VA.

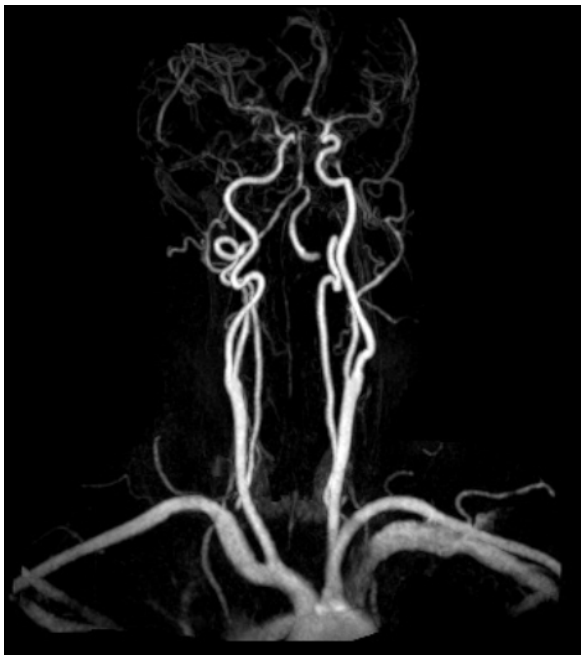

Figure 28. Contrast-enhanced MRI angiogram (MRA) showing flowing blood. This view is through the brain and carotid arteries. *Courtesy of University of Iowa Carver College of Medicine.*

Contrast agents can be used in MRI to enhance the brightness of particular body tissues, including intravascular blood. Recall that because gadolinium-based contrast agents work by shortening T1 relaxation times, imaging techniques utilizing contrast agents almost always employ T1-weighted pulse sequences, such as rapid gradient-echo imaging techniques. Time-of-flight imaging for creating MR angiograms is the primary example. MR contrast agents are also used for perfusion imaging in the brain and heart.

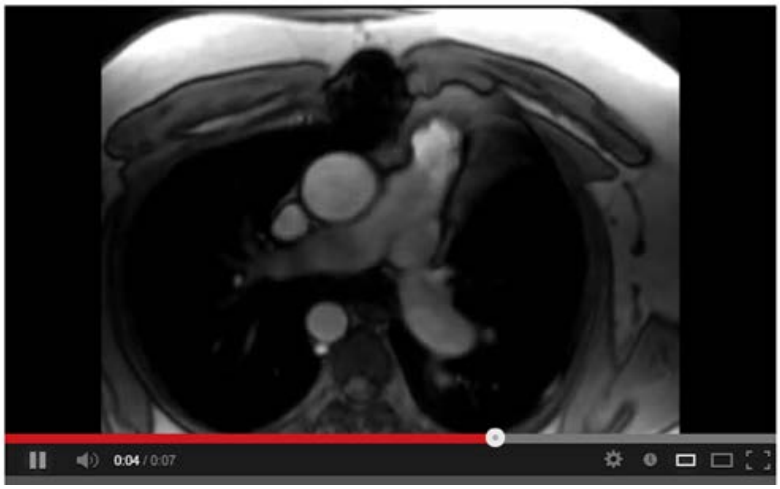

Figure 29 MOVIE. Multiphase, phase contrast sequence demonstrating flow through the aortic valve. The brightness in the phase contrast images relate to the direction and velocity of blood flow. Available at: YouTube.com/ICPMEducation. *Courtesy Fairfax Radiological Consultants, Fairfax, VA.*

Phase Contrast

Another method for demonstrating flowing blood is called **phase contrast** (PC). This principle of phase contrast imaging is similar to TOF imaging in that moving tissue generates a different signal compared to stationary tissue. In phase contrast imaging, however, extra flow-encoding gradients are turned on after excitation. These gradients act like phase-encoding gradients, causing the nuclei that are moving to have a different phase than for the stationary nuclei. The image can be subtracted from an image without flow-encoding gradients to generate a map of only flowing blood (**Figure 29**).

Compared to time-of-flight imaging, phase-contrast imaging is slower, and the images are of lower resolution. However, a major advantage of phase-contrast imaging is that both direction *and* velocity of flow can be measured. This is important for many applications such as cardiac imaging. Both time-of-flight and phase-contrast imaging are useful MRI tools for examining blood vessels throughout the body.

Diffusion-weighted Imaging

Diffusion is the microscopic and random motion of molecules such as water. Diffusion of water occurs in all body tissues but because of the structure and arrangement of cells and other materials in the body, the actual motion is not truly random. As a result, the rate and direction of diffusion in body tissues provides diagnostic information about the tissue.

Diffusion-weighted imaging (DWI) works by turning on very strong gradients after the initial excitation. As with phase contrast imaging, this makes moving protons have a different (lower) signal than stationary protons. Areas of high diffusion will appear darker on DWI and areas with reduced diffusion will appear brighter.

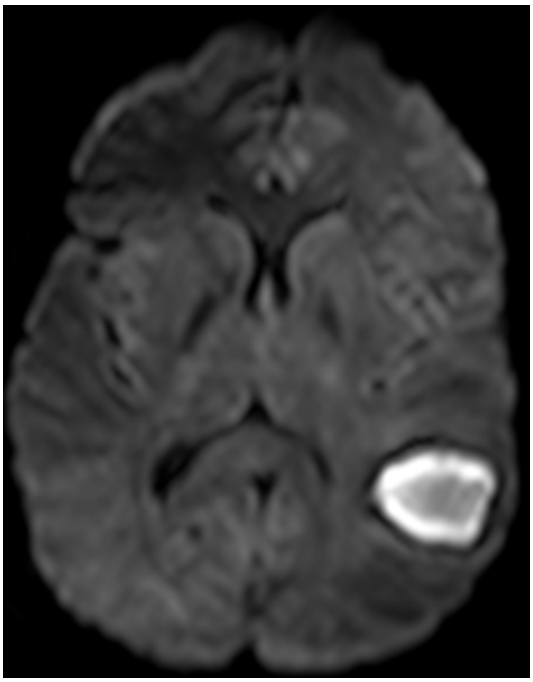

Figure 30. Axial brain diffusion-weighted image (DWI) of resolving hematoma. *Courtesy of Jacqueline Bello, MD, FACR, Montefiore Medical Center.*

This has proven valuable in imaging the brain of stroke patients or patients with cerebral hemorrhage where diffusion is significantly reduced in the affected area of the brain and results in bright signal (**Figure 30)**. DWI can be performed with spin-echo or gradient-echo pulse sequences but is most commonly used with rapid echo-planar imaging techniques.

An extension of DWI, **diffusion tensor imaging** (DTI) is an MRI-based neuroimaging technique that maps the three-dimensional diffusion of water molecules throughout the brain. Since diffusion of water in the brain primarily follows the directions of its white matter tracts, DTI makes it possible to visualize the location, orientation, and connections of those tracts with fiber tracking, also called **tractography**, to follow the path of fibers along their whole length. DTI measures both the amount and direction of diffusion. This information can create 3D models of the neural pathways (**Figure 31**).

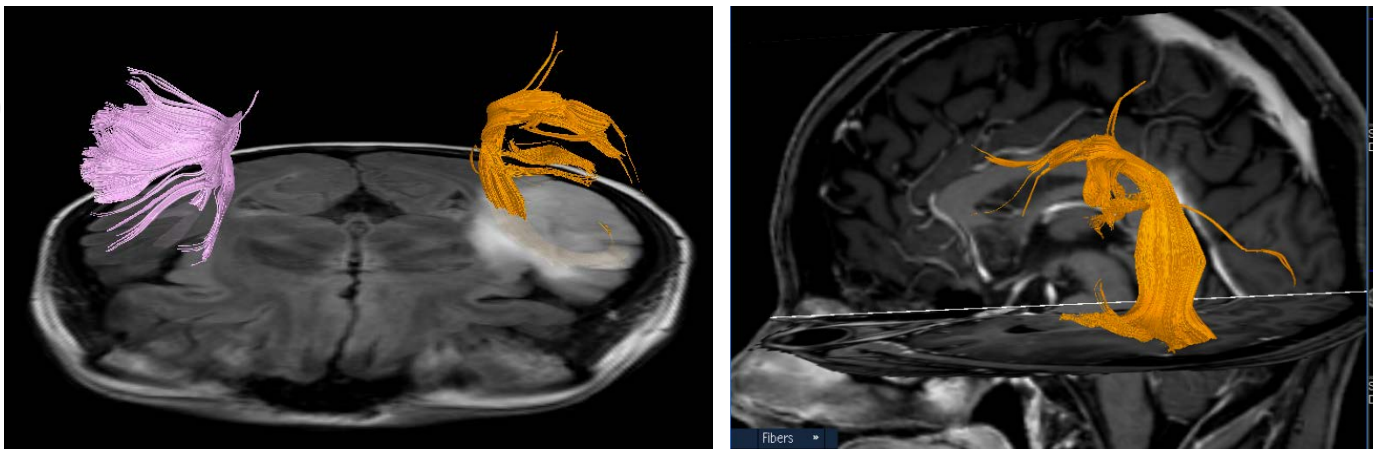

Figure 31. Examples of diffusion tensor imaging (DTI) generated by tractography. The fiber tracts of the brain are visualized, *Courtesy of Jacqueline Bello, MD, FACR, Montefiore Medical Center.*

Pulse Sequences

We know that the two fundamental types of pulse sequences used in MRI are spin-echo and gradient-echo. Most other pulse sequences used in MRI are variations of one of these. Changes may include additional pulses like inversion recovery or acquisition of additional data during the interval of a single repetition.

Variations of Spin-Echo-based Pulse Sequences

Fast or turbo Spin-Echo

The original spin-echo pulse sequence described earlier produces high-quality images but requires a long time to acquire the data for high-resolution images. An updated version of this pulse sequence called **fast spin-echo** (FSE; GE) or **turbo spin-echo** (TSE; Siemens and Philips) reduces the time required for a complete image by a factor of ten or greater and is now almost universally used for T1- and T2-weighted imaging.

Recall that in the standard spin-echo pulse sequence, one line of data in *k*-space is taken after each excitation. In FSE/TSE, additional refocusing pulses are added after the initial 90° excitation, generating additional echoes that fill in additional lines of *k*-space. The number of echoes recorded during each repetition is called the **turbo factor** or the **echo train length**. **Figure 32** illustrates this sequence of pulses. The amount of time required for a complete image is reduced by a factor equal to the turbo factor, which shortens the imaging time considerably with very little loss in image quality or contrast.

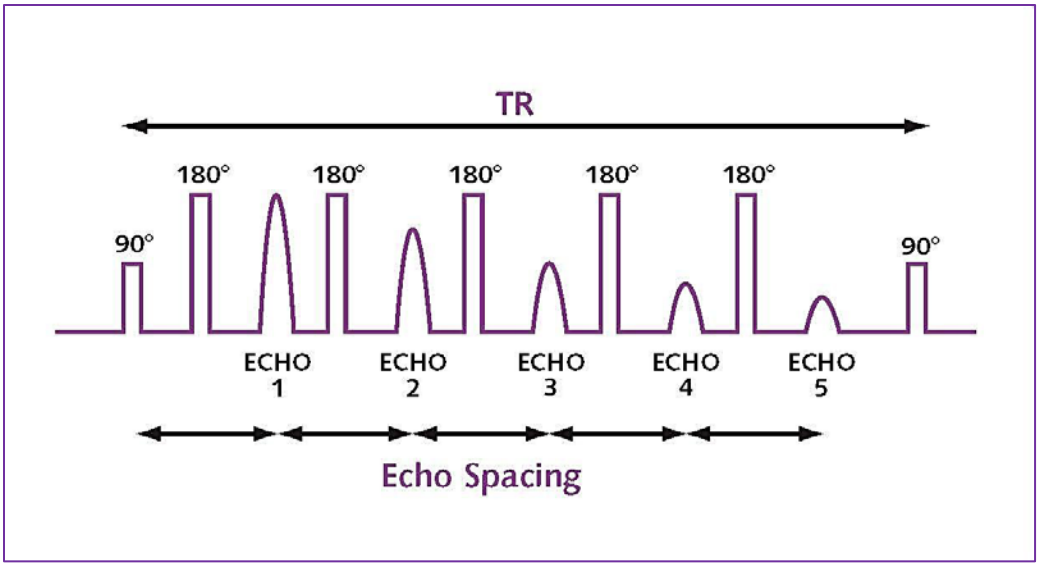

Figure 32. Turbo factor or echo train length of 5 echoes in a FSE/TSE pulse sequence.

Variations of Gradient-Echo-based Pulse Sequences

Echo Planar Imaging

We already know that the gradient-echo pulse sequence described earlier creates images faster than the standard spin-echo pulse sequence. Still, for some applications there is a need to generate images as quickly as possible. With even faster MRI using gradient-echo imaging, it is possible to acquire many complete images in less than a second. This means that a "movie" of motion or a dynamic process is possible with MRI, adding another dimension to the information provided to clinicians.

Echo planar imaging (EPI) is the fastest of the gradient-echo-based scanning protocols. Similar to fast spin-echo, multiple lines of *k*-space can be acquired after each excitation, that is, over the duration of each TR. In EPI, the number of lines of data obtained varies from 8-16 up to all of the lines required for a single image. A complete image can be acquired in tenths of a second. **Figure 33** illustrates part of the pulse sequence diagram for an echo planar acquisition.

However, these very short imaging times come at some cost. The resolution and image quality of echo planar images are usually much lower than standard gradient-echo or spin-echo images. In addition, the hardware requirements needed for echo planar imaging are demanding, and not all scanners have hardware optimized for this application. Nevertheless, when a very short scan time or a rapid sequence movie of a dynamic process is needed, EPI makes this possible.

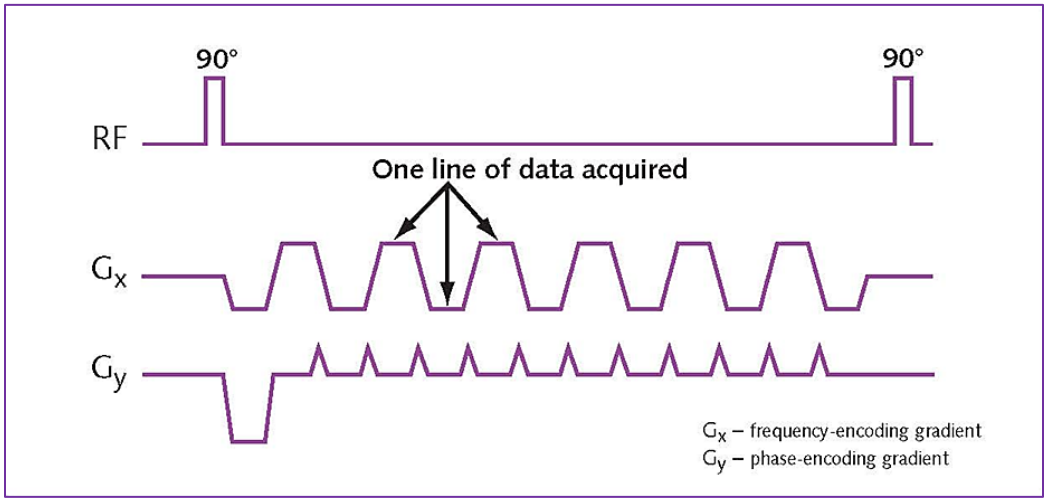

Figure 33. Pulse sequence showing echo planar acquisition.

RADIAL IMAGING

Another gradient-echo-based method, **radial imaging**, is becoming more common. As with all gradient-echo sequences, there is no refocusing pulse. The difference in radial imaging is that the data in *k*-space are not taken line-by-line but in a "fan" pattern from the center of *k*-space towards its edge.

One of the advantages of radial imaging is that echo time can be extremely short. This makes tissues with a short T2 visible that are not seen with other imaging protocols. Special reconstruction techniques can create rapid movies in applications like cardiac imaging, taking advantage of the fan-like data pattern. Correction for motion artifacts can also be done with radial imaging, using techniques such as BLADE (Siemens), PROPELLER (GE), and MultiVane (Philips). Image reconstruction for radial MRI is more complicated than for standard MRI sequences, but with the availability of robust computer systems, this is no longer a limitation.

Steady-state Free Precession and Fast Imaging with Steady-state Precession

Another family of pulse sequences gaining increased use is **steady-state free precession** (SSFP) (FISP, Siemens; FIESTA, GE; Balanced FFE, Philips). Because SSFP imaging lacks a refocusing excitation pulse, it is most closely related to gradient-echo imaging. The difference is that with SSFP, all of the gradients used are symmetric, which helps preserve as much signal as possible throughout the acquisition. The repetition times used are very short, yielding more rapid imaging than conventional gradient-echo and potentially making the acquired images less sensitive to motion from breathing.

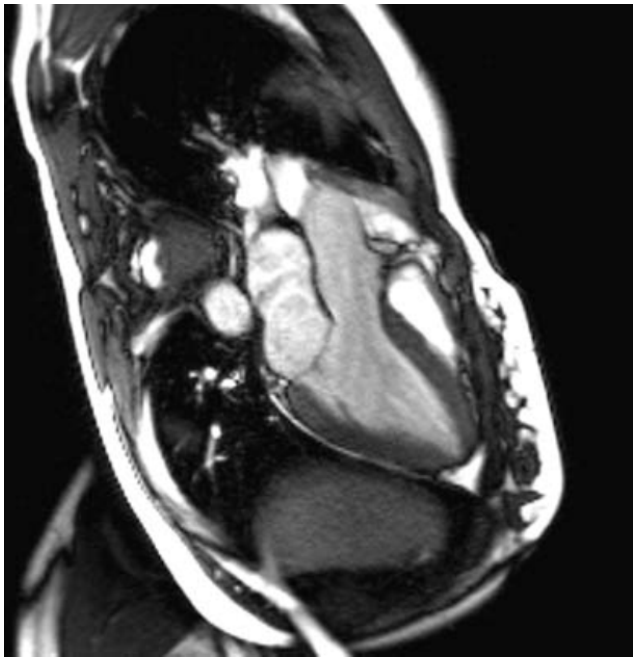

Figure 34. Example of SSFP of the heart. *Courtesy of University of Iowa Carver College of Medicine.*

With SSFP imaging, image contrast is not purely based on T1, T2, or proton density but is a combination of these factors. It is most closely related to the ratio of T2/T1. With SSFP, tissues demonstrate different intensity than on T1- or T2-weighted imaging alone. The mixed T1/T2 contrast of SSFP and its reduced sensitivity to motion make SSFP well-suited to cardiac and body imaging applications (**Figure 34**).

Summary

We have introduced several commonly used pulse sequences in MRI, demonstrating the tremendous versatility of this imaging modality. The complete list of MR pulse sequences and types of image contrast is much longer because specialized techniques can be tailored to many areas of the body. Examples include perfusion imaging, functional imaging, chemical-shift imaging, and spectroscopy.

The utility of any MRI protocol depends on the gathering of high-quality images. MRI is a mature field, so standard protocols for common exams are widely available. However, care must be taken by the scan operator to ensure high quality. We will next address factors affecting image quality. Image quality involves tradeoffs in scan time, resolution, and contrast. Commonly observed artifacts are discussed, in addition to how they can be averted or corrected.

NOTES

POINTS for PRACTICE

1. The resonant frequency of fat protons is slightly lower than the Larmor frequency of water protons. What is this difference called?
This difference is known as chemical shift. The chemical properties of fat cause a shift in the frequency at which it resonates compared to other substances. Because of this difference, it is possible to make an excitation pulse at the resonant frequency of only fat or only water and not of other tissues.

2. How does inversion recovery change the appearance of some tissues?
Inversion recovery adds an additional excitation pulse prior to the rest of an ordinary spin-echo or gradient-echo pulse sequence. A 180°, or inversion, pulse flips the magnetization of the nuclei to point in the opposite direction, and the spins start to recover back towards their equilibrium state at a rate that depends on T1 relaxation time. With proper image acquisition timing, inversion recovery can be used to null the signal from a particular tissue.

3. Define FLAIR and STIR.
FLAIR (fluid-attenuated inversion recovery) is the use of an inversion pulse, followed by a delay until the halfway point of the relaxation of a fluid while other tissues recover their signal. This will show a significant contrast between the two, for example, nulling the appearance of CSF vs surrounding tissues. STIR (short T1 inversion recovery) works the same way as FLAIR but involves the use of a short inversion time (TI) to differentiate, for example, fat from surrounding tissues.

4. Which imaging acquisition methods are widely used to display blood flow?
Time-of-flight (TOF) imaging is often used to display flowing blood. The purpose of TOF image acquisition is to make stationary tissue appear dark on the image, while making moving tissues such as blood show up bright. The resulting image reveals only flowing blood, which provides an outline of the blood vessels. Phase contrast is similar to TOF but is slower and creates lower resolution images. However, a major advantage of phase contrast imaging is that both direction and velocity of flow can be measured.

5. Name a more efficient version of spin-echo and explain why it is more efficient.
Turbo spin-echo (TSE) or fast spin-echo (FSE) utilizes additional refocusing pulses, generating more echoes from each excitation that fill in more lines of *k*-space. The amount of time required to produce an image with this technique is much shorter.

6. Which is the fastest of the gradient-echo-based pulse sequences?
Echo planar imaging (EPI) is the fastest of the gradient-echo-based pulse sequences. Multiple lines of *k*-space are acquired after each excitation but at a much faster rate, although resolution and image quality may be lower than that of standard images.

7. What type of imaging uses a "fanning" type sequence?
Radial imaging acquires the data in *k*-space not line-by-line but in a "fan" pattern from the center of *k*-space towards the edge. This technique makes images with a short T2 visible, and it has special reconstruction techniques that can produce rapid movies.

IMAGING PARAMETERS and IMAGE QUALITY

MRI is a flexible imaging modality capable of generating a wide range of useful information. The selection of a particular MR acquisition technique depends on the information required to answer the clinical question. A good deal of skill on the part of the scan operator is required to ensure that not only is the needed information obtained but that the images are of high quality.

The technologist needs to be well-versed in the effects of the many MRI parameters that can be manipulated during an exam. It is essential that the technologist is able to assess image quality and appearance and to solve problems that arise to ensure that the needed information is acquired efficiently and accurately.

We will next define some measures of quality in imaging. We will also describe common problems and image artifacts that frequently arise in the clinical setting and offer ideas as to how artifacts may be eliminated or reduced.

POINTS for PRACTICE

1. What are the three general characteristics that describe the quality of an MR image?

2. What are some parameters that affect image quality that are not scan operator-controlled?

3. Name some parameters that can affect image quality that are scan operator-controlled.

4. Provide some examples of artifact.

5. As the scan operator, you must decide what variables are essential for obtaining the best images. What are some of the trade-offs needed to produce consistent quality?

6. Increasing matrix size and decreasing FOV both alter image quality criteria in the same way. What are the criteria? What accounts for the change?

Measures of Image Quality

Let's review the three general characteristics that describe the quality of an MR image: **spatial resolution**, **signal-to-noise ratio** (SNR), and **image contrast**.

Spatial Resolution

Spatial resolution defines the ability to distinguish two structures an arbitrarily small distance from one another as separate, that is, how much structural detail is captured in an image. Images with higher spatial resolution permit smaller structures to be seen and distinguished, resulting in crisper, sharper images. Images with low spatial resolution appear blurry with indistinct content, making it difficult to clearly identify structures.

Spatial resolution depends on the matrix size and field of view acquired, that is, the number of picture elements or **pixels** acquired in each direction of the image. In a larger matrix, there are more pixels for a given area, so more detail is potentially seen. However, a large matrix size does not guarantee that an image will be sharp, since other effects such as motion can affect the image. The matrix size and field of view determine the greatest resolution achievable in the image.

Signal-to-Noise Ratio

In addition to the desired signal used to create the image, random and unwanted variations — **noise** — will be added to the signal. The signal-to-noise ratio relates to the intensity of desired signal from body tissues relative to background noise. A high-quality image will have a high SNR. Low SNR images look grainy, and some anatomic structures may be difficult to visualize (**Figure 35**). SNR can be influenced by several protocol parameters under the control of the operator, such as FOV and matrix size, TR and TE, and **number of excitations** or **number of averages** (NEX/NSA).

Image Contrast

Image contrast is the difference in brightness between two structures or regions within an image. In an MR image, this reflects the difference in the signal received from the different tissues. This may be dependent on the number of hydrogen nuclei (proton density), the T1 and T2 relaxation parameters of the tissues, or any of several other factors previously discussed. In imaging terms, high signal intensity corresponds to bright areas of the image and low signal intensity to dark areas.

As with SNR, a measure of contrast compared to the random variations in intensity due to noise can be measured. The **contrast-to-noise ratio** (CNR) is the difference in intensity between two tissues of interest relative to the noise level. For example, if two tissues appear equally bright on an MR image, they may both have a high SNR but because they look similar, they have a low CNR. Even if the image is of high quality in terms of SNR, it may not be useful if different tissues cannot be distinguished. Speed, detail, and quality represent three interchangeable variables in image acquisition, where one variable is traded off in order to enhance the other two. For instance, a detailed, high-quality image will require a long acquisition time. Conversely, if a detailed (high-resolution) image is acquired rapidly, it will have low SNR**.**

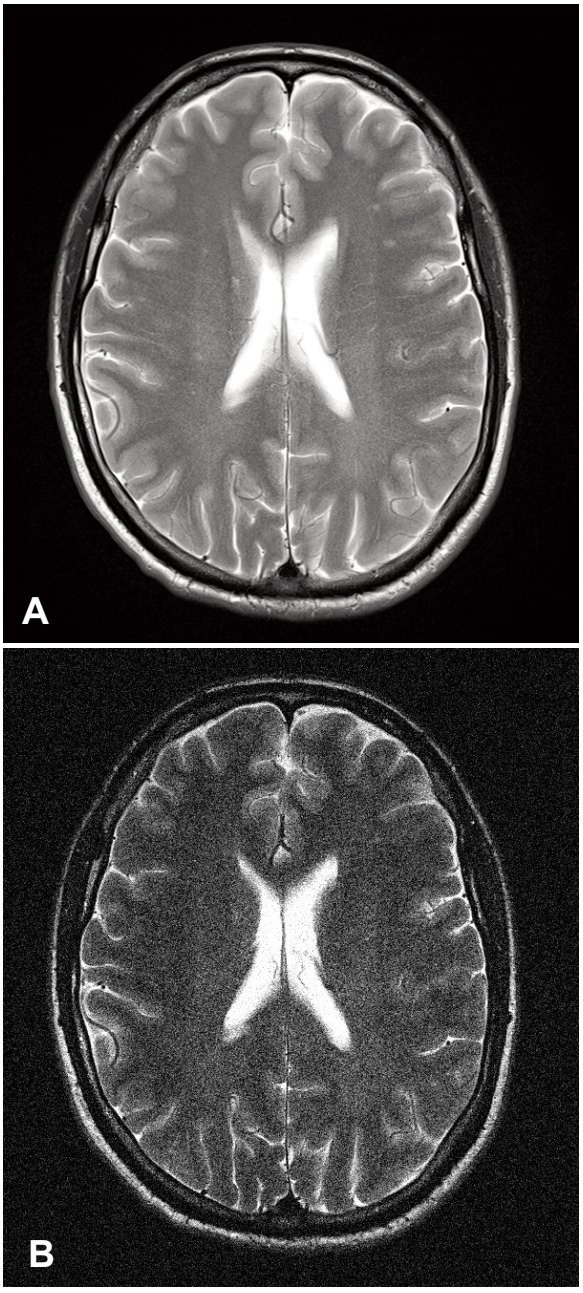

Figure 35. Axial T2-weighted image of the brain. (A) Example of high SNR with no artifact. (B) Example of low SNR. *Courtesy of University of Iowa Carver College of Medicine.*

All three of these elements — spatial resolution, SNR, and image contrast — are interdependent. However, a deficiency in any one of these three elements cannot be compensated for by the other two. For example, even an image with high SNR cannot overcome a lack of detail and likewise, an image with high anatomical detail is unusable if image contrast is poor.

Speed, detail, and quality represent three interchangeable variables in image acquisition, where one variable is traded off in order to enhance the other two.

Elements that Influence Image Quality

Many characteristics influence the quality of an MRI scan. Some of these traits relate to the tissue and are not under the control of the operator. Others relate to the imaging parameters selected during the acquisition that can be adjusted to create the best trade-off of quality, detail, and speed and are under the operator's control.

Non-operator Controlled Image Acquisition Parameters

What are some of the physical characteristics of scanned tissues that influence image quality and are outside the control of the scan operator? While the scan operator cannot control these physical characteristics, changes to other scan parameters may alter their influence on signal intensity in the final image.

Spin Density and Relaxation

The MR signal comes from the sum total of signal generated by each excited hydrogen nucleus. Since the brightness or intensity in the image depends on the signal at each location, a higher density of hydrogen protons, or spins, creates more signal and therefore a brighter region on the image.

In basic spin-echo and gradient-echo pulse sequences, higher signal intensity is seen in tissue with a short T1 relaxation or a long T2 relaxation. However, recall that advanced pulse sequences make the dependence on T1 and T2 relaxation somewhat more complicated. For example, inversion recovery imaging may completely change the relationship between T1 and signal intensity.

Gadolinium-based Contrast Agents

We have noted that MR GBCAs work by shortening the T1 relaxation time in areas reached by the contrast agent. The effect of contrast agents on the image is the same as for any tissue with short T1 relaxation. This effect creates higher signal and brighter image intensity in the pulse sequences in which the GBCA is used. This is especially true since the T1 relaxation time of the contrast agent is much shorter than the relaxation times of body tissues in the absence of contrast.

Flow and Motion

Flow and motion have considerable effects on image intensity and image quality, but these effects are complex. Unless the pulse sequence selected is designed to suppress or enhance changes due to flow and motion, these parameters typically have negative effects on image quality by reducing signal or causing blurring or other artifacts. Among other factors, the degree of influence depends on the rate and pattern of flow or motion, slice thickness and orientation relative to the motion, and the gradients and pulse sequences chosen.

Instrument (Scanner) Properties

Another set of factors that influence quality is the system hardware and associated coils. The effect of hardware and coils on image quality cannot be changed at the time of the scan but should be understood by the technologist. For sites with multiple scanners and coils, system hardware factors may influence how patients are directed to particular scanners to achieve the best possible image for specific clinical questions.

B_0 field strength: low field, 1.5T and 3.0T

Recall that B_0 field strength results in greater signal from tissue than from the same tissue imaged at lower field strength. All other factors being equal, a high field magnet produces a higher quality image than one of lower field strength. However, field strength also affects other factors that influence image quality. For example, the T1 relaxation time for a given tissue increases at higher field strengths. RF heating becomes more of an issue at higher field strengths, which may require protocol changes such as longer TR times, reduced flip angles, or reduced number of slices. The availability of RF coils may also vary depending on field strength so that practically speaking, a lower field strength with an appropriate coil may be preferable to a high-field system with a less optimal coil. At present, neurological and musculoskeletal exams can best take advantage of higher field strength scanners such as 3T, while body and cardiac exams are still more commonly performed at 1.5T.

B_0 field homogeneity

In MR imaging, the main magnetic field is designed to be as uniform and homogeneous as possible over the volume scanned. Still, each magnet has some imperfections, or **inhomogeneity**, within the magnetic field over this volume. The degree of inhomogeneity depends on the type of magnet — resistive, permanent, or superconducting — and the shape of the design. Additional hardware like shim coils is often included to adjust for such imperfections, and **shimming** to “smooth out” the magnetic field is routinely performed for each patient exam. These steps are important in preventing artifacts and geometric distortion associated with magnetic field inhomogeneity. Again, field strength is also a factor, with 1.5T magnets generally having better homogeneity than 3T systems.

Gradient strength, speed, and accuracy

Recall that the three gradient coils required for each of the principle orientations surround the inside of the bore.

Gradient coils are engineered to perform rapidly and accurately over the scan volume but still have some imperfect performance characteristics. The speed and strength of gradients determine the limits of many MRI parameters such as TE and bandwidth. Inaccuracies in the gradient can also cause image artifacts such as geometric distortion, or **ghosting**, especially for rapid scans like echo planar imaging that push the gradient system to its limits.

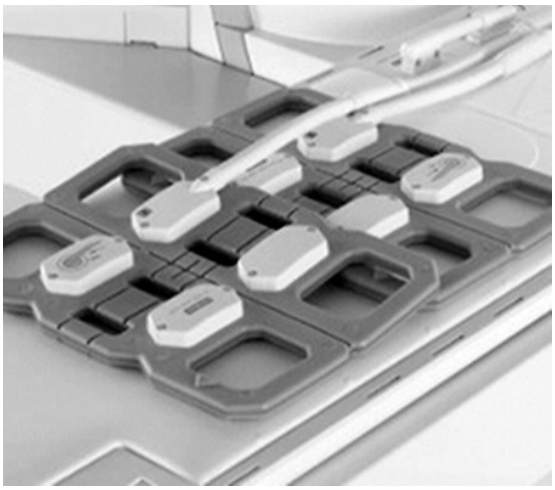

Figure 36. 16-channel phased-array coil, primarily used for cardiac and body applications. *Courtesy of GE Healthcare*

RF coils

Most MRI facilities include a selection of coils for scanning different anatomic areas of the body. Basically, smaller coils image only smaller regions of the body but have higher image quality because they pick up less noise, yielding improved SNR. The choice of RF coil for a particular exam is usually the smallest coil that covers pertinent anatomy. Phased-array coils incorporate multiple coil elements as part of a single unit to take advantage of the higher quality or faster scans that can be achieved using techniques such as parallel imaging. As with magnet and gradient designs, RF coils have their own performance characteristics such as uniformity of RF energy, making the quality of each coil unique (**Figures 36-38**).

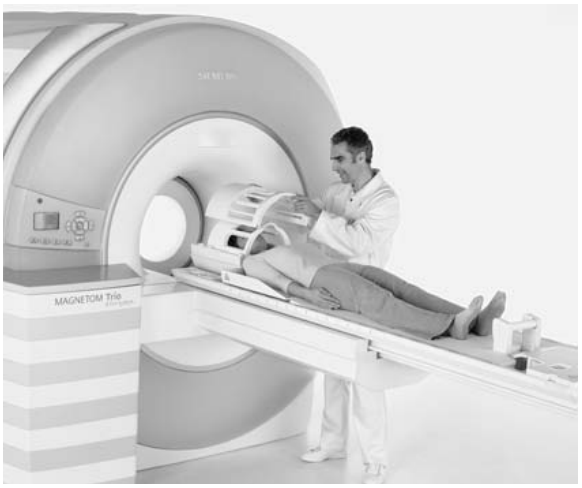

Figure 37. Example of placement of a head coil. *Courtesy of Siemens.*

Other factors

Several other factors of the MR imaging process affect image quality. These factors are dependent on the specifications of the scanner manufacturer. Examples include conversion speed of the MR signal into a computer-readable form, amount of memory, and image reconstruction operations. Familiarity with the limits of a particular system is important for ensuring high-quality diagnostic images.

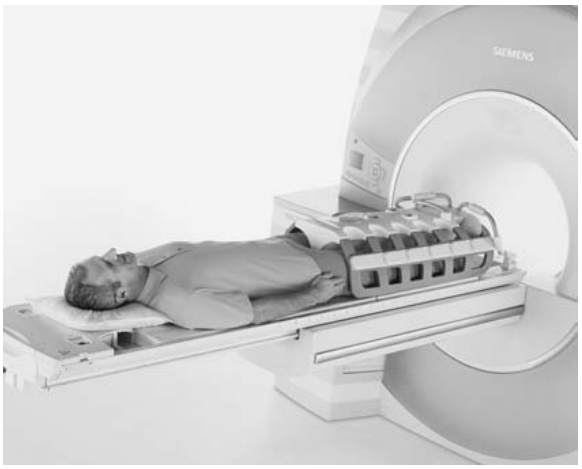

Figure 38. Example of placement of a runoff coil used for imaging blood vessels of the legs. *Courtesy of Siemens.*

Operator-Controlled Image Acquisition Parameters

Many of the parameters that influence image quality *are* under the control of the scan operator and a thorough understanding of how the operator can impact image quality when building or modifying protocols is essential.

Slice Location and Slice Orientation

The location of the selected slices affects image quality. The main magnetic field and gradients are designed to be most homogeneous at the center of the magnet. As a result, the homogeneity at the outer margins of the scan region may be inferior to that at the center, potentially creating distortion and artifacts associated with inhomogeneity. Ideally the patient should be positioned so that the anatomy being scanned is near the center of the magnet bore, though this is not always possible for reasons of patient size, positioning of the anatomy of interest, and comfort. Most modern scanners allow the scan table to be automatically positioned such that the imaged volume is at the center for each scan in a protocol, that is, at **isocenter.**

The orientation of the slices must be selected with care by the scan operator. To avoid aliasing, a common artifact discussed later, the field of view in at least one dimension must be large enough to cover all of the tissue in the imaging plane. If the orientation is such that the FOV must be very large along the phase-encoding direction, the scan time may become long or the resolution reduced. The choice of slice orientation and phase-encoding direction is important for efficiently covering the desired anatomy.

Slice Thickness and Spacing

The selection of slice thickness influences both the resolution and image quality in terms of SNR. The minimum slice thickness achieved depends on the capabilities of the gradient system and the duration of the RF pulse used. As slices are made thinner, the resolution of the image improves. However, the smaller volume of tissue included in the slice means that the total signal is reduced, with a corresponding reduction in the SNR of the image. Confronted with another trade-off, slice thickness should be chosen at the minimum thickness that provides sufficient image quality and SNR.

It must also be recognized that slice selection excitation is not perfect. There will be a small amount of tissue outside the selected slice that will be excited and therefore generate signal in the image. This phenomenon is called slice **crosstalk** and causes further artifacts in an acquisition with multiple slices as the excitation for one slice will crosstalk in the next slice and thereby affect its magnetization. When the excitation for the next slice is performed, the magnetization may be partly saturated and will have reduced signal. One solution is to place gaps between the slices so that there is less crosstalk. Another is to arrange the order of the slice acquisitions so that slices that are located next to each other are excited as far apart in time as possible, known as slice **interleaving** (**Figure 40**).

Resolution/Matrix Size

The matrix size of an image is simply the number of pixels acquired in each dimension of the image. MR images typically have a 256 × 256 matrix size. Resolution, measured in millimeters per pixel (mm/pixel) in each direction, is the field of view divided by the matrix size for each direction. The resolution represents the volume of material included in just one pixel of the image. If the matrix size is increased and the field of view stays the same, the resolution in mm/pixel will be smaller, meaning there is less tissue to provide signal. As a result, SNR is reduced in proportion to the increase in the matrix size. This explains why higher resolution images (those having smaller mm/pixel sizes) demonstrate reduced SNR. If the pixel size is too small or the matrix size too large, the image may become noisy and have a grainy appearance.

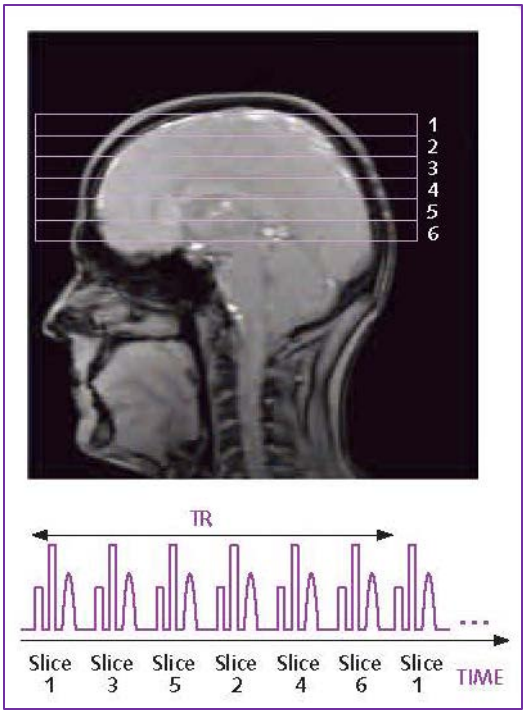

Figure 40. Slice interleaving. Adjacent slices are acquired farther apart in time to reduce crosstalk artifact. *Courtesy of University of Iowa Carver College of Medicine.*

Field of View

The field of view affects image quality and SNR similarly to matrix size. If the matrix size remains the same and the FOV made smaller, the resolution in mm/pixel will be lower and thus reduce the SNR of the image. A larger FOV with the same matrix size will give a higher SNR, since each pixel covers a larger volume (more mm/pixel) and therefore contains more protons. In practice, FOV is selected to cover the anatomy of interest, while matrix size is adjusted to set the appropriate resolution and scan time. The FOV must be large enough so that no aliasing artifact results.

Averages

For any set of imaging parameters, the SNR is increased by averaging, which simply acquires the same image multiple times and then averages the images. Recall that the number of times the same image is acquired is referred to as the number of excitations or number of signal averages (NEX/NSA).

The signal from each acquisition should be the same in each image but noise is random, and noise from one acquisition is partly canceled out in the next acquisition. This means that when the signal is the same, noise is reduced and SNR is higher. Numerically, the SNR increases as the square root of the number of averages, so an image acquired with two averages (NEX = 2) has an SNR that is approximately 1.4 times greater.

The obvious limitation of increasing the number of averages is that scan time increases; therefore, the number of averages must be kept low enough that scan time remains reasonable.

Bandwidth

The bandwidth of an acquisition relates to how fast the system records the signal after each excitation. It may be indicated in kilohertz (kHz) or in hertz per pixel (Hz/pixel), depending on the manufacturer. A higher bandwidth corresponds to a faster rate at which the signal is recorded. Higher bandwidths may allow the TE to be shorter, as well. However, the amount of noise goes up as the bandwidth is increased. This means that increasing the bandwidth decreases the SNR and image quality. Increasing bandwidth may help reduce some types of artifacts, such as those related to metal implants, for example, hip and knee joint replacements.

Flip Angle

In a gradient-echo pulse sequence, the flip angle is adjusted to affect image quality and SNR. It may change the amount of T1 contrast present in the image. As a result, the relationship between the flip angle and SNR is complicated. We have already described how the flip angle is chosen to maximize signal for a particular combination of tissue type and selected TR. This means that the flip angle selected provides the highest SNR for tissue type and selected TR; changing either parameter would decrease SNR for that tissue. If the TR is lowered, then the flip angle also should be lowered to retain the best image quality, although SNR is still lower. Because SNR and contrast are both tissue- and TR-dependent, the optimal flip angle depends on the type of scan and anatomy imaged.

Filtering and Post-processing

The scan console provides additional options for acquiring and viewing images designed to improve their appearance. These may include adjustments in contrast and brightness (also called **window and level**) or filters to reduce noise or smooth the images. Brightness and contrast adjustments assist in making desired features more apparent but do not change SNR or other image properties. Image smoothing operations may reduce noise in the image but also result in blurring, decreasing the resolution of the images. In either case, the original source images acquired must be retained so that complete data are available to review.

Artifacts

Even with a well-tuned system and imaging protocols that provide excellent image quality, things can go wrong during a scan, resulting in poor image quality or image artifacts. We will next discuss the most common imaging artifacts and ways to reduce or eliminate them.

ALIASING

When planning an MRI acquisition, it is important that all of the tissue in the selected slice generates signal and is included in the imaging field of view. **Aliasing**, also known as wrap, occurs when the field of view selected for the scan is smaller than the size of the anatomy in the imaging plane. Because the tissue outside of the FOV still experiences the excitation pulse and generates signal, it appears in the image but is "wrapped" to the opposite side of the image (**Figure 41**).

On most scanners, aliasing is only a problem in the phase-encoding direction or the slice-select direction of a 3D acquisition. One solution is to switch the frequency and phase directions so that the narrower dimension of the anatomy is in the phase-encoding direction. Another solution is to increase the FOV along the phase-encoding direction, resulting in decreased resolution for the same scan time or a longer scan time with the same resolution. The latter strategy is referred to as **oversampling**.

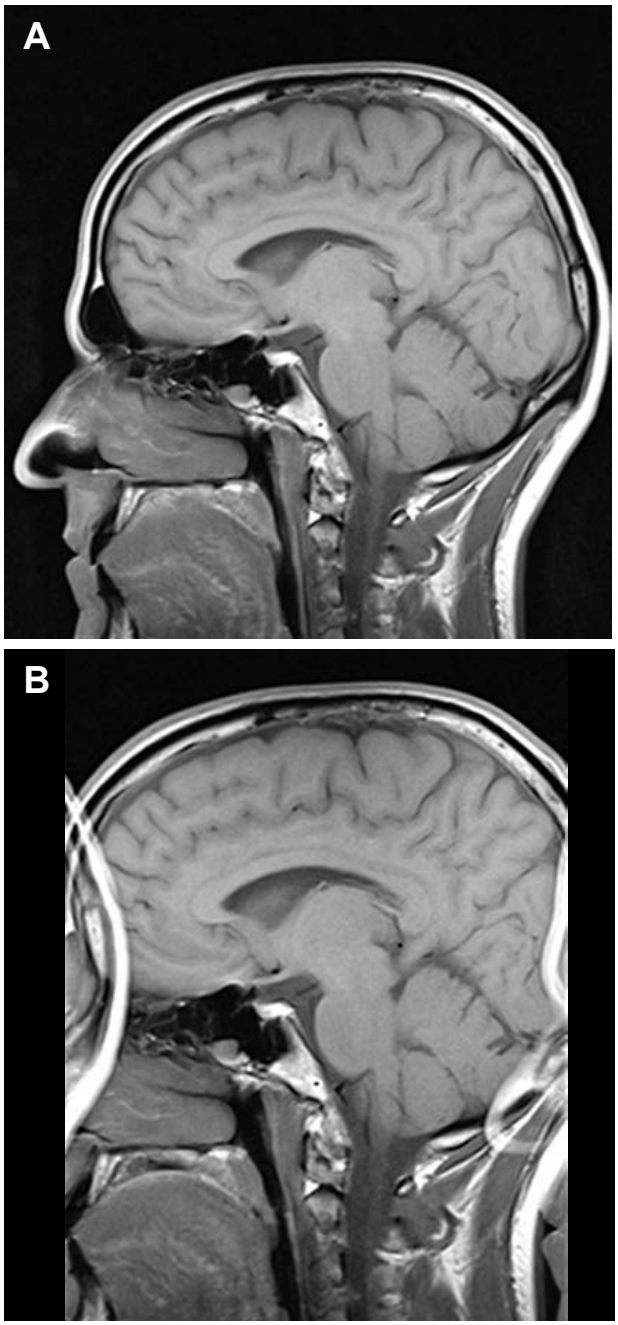

Figure 41. Sagittal brain image. (A) Good quality image without aliasing. (B) Poor quality image with aliasing. *Courtesy of University of Iowa Carver College of Medicine.*

MOTION

Since a typical image acquisition in MRI lasts a few seconds to a few minutes, it is important that the patient be as still as possible. If there is patient movement during the scan, the resulting image may be corrupted, similar to taking a photograph with a long exposure time (**Figure 42**).

In MRI, motion may make the image appear blurry, or it may introduce ghosting artifacts that appear along the phase-encoding direction. Any type of motion can cause these artifacts. Frequently patients become uncomfortable and shift during the scan, and proper placement of pillows and pads helps reduce this. If a patient does move, the only solution is to make the patient more comfortable and repeat the scan.

Regular breathing motion causes artifacts when imaging the chest and abdomen. Scans of these areas utilize protocols that require patients to hold their breath. Most scanners also monitor breathing and thereby can acquire data during exhalation for long scans or in patients who are not able to reliably hold their breath. This process is known as **respiratory gating** or **respiratory triggering**.

Similarly, the motion of flowing blood also introduces artifacts into the image. Here, the addition of scan options such as the application of specifically shaped gradient pulses called **flow compensation** or synchronizing the scan to the heartbeat, called **cardiac gating**, can help eliminate these problems. As always, there are trade-offs with these techniques in the form of longer scan times or longer echo times.

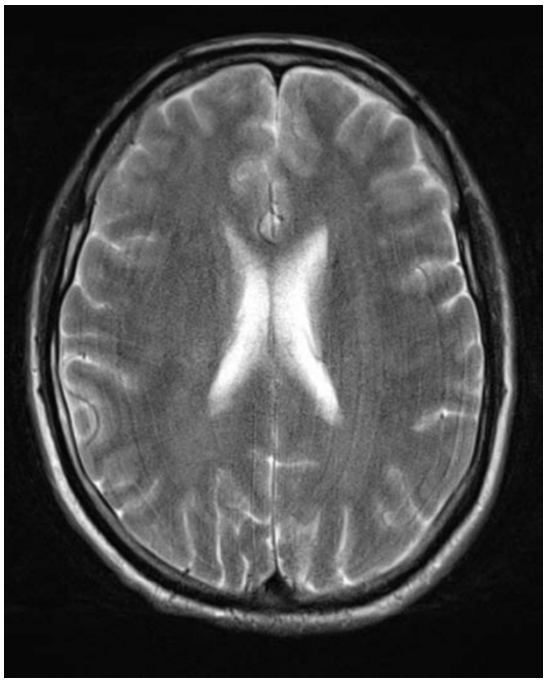

Figure 42. Axial brain image showing motion artifact. *Courtesy of University of Iowa Carver College of Medicine.*

Chemical Shift

Tissues composed mostly of fat create artifacts on MRI. Because of the characteristics of fat, the magnetic field experienced by the hydrogen nuclei of fat is slightly different from that in other tissues and therefore the nuclei spin at a slower frequency. Remember that the frequency of spinning is used to determine where the tissue is within the image. Fat appears to be slightly shifted along the frequency direction relative to other tissues in the image. The amount of shift is related to the bandwidth, and higher bandwidth yields reduced shift. Recall that fat artifact is called chemical shift because it arises from the chemical properties of fat.

Chemical shift artifacts may mimic pathology, so it is important to distinguish whether imaging features are real or artifactual.

Chemical shift displaces the appearance of fat in the image, causing the fat to overlap with other tissues on one side while leaving a gap on the opposite side. This creates a bright band where the overlap occurs and dark band where there is signal void. These artifacts may mimic pathology, so it is important to distinguish whether these features are real or artifactual. A solution to this challenge is to use some form of fat suppression such as fat saturation which, as discussed earlier, completely eliminates the signal generated by fat. The bandwidth of the scan can also be increased, which reduces the amount of chemical shift.

SUSCEPTIBILITY

Other factors may cause the magnetic field to be altered or distorted in specific parts of the body. For example, metallic implants such as orthopaedic screws cause changes in the nearby magnetic field. Tissues adjacent to pockets of air such as the paranasal sinuses can alter the magnetic field. These alterations are caused by changes in **magnetic susceptibility** that distort the normally constant field of the main magnet.

These susceptibility changes will cause the nearby magnetic field is inhomogeneous, resulting in either distortion of the image or complete “black holes” where there is no signal at all. It is important to recognize whether or not these susceptibility changes represent artifact so that the correct diagnosis can be made.

To minimize susceptibility artifact, the bandwidth of the scan is usually set very high, and extra steps like shimming may be performed to adjust the magnetic fields to make them as uniform as possible. In the case of metal implants, such as those used for hip and knee joint replacements, metal-artifact-reduction (MARS) techniques can be employed to reduce metal artifacts.

In addition to increasing bandwidth and the use of shimming, additional steps can be taken to minimize susceptibility artifact:

- using a lower field strength scanner
- avoiding gradient-echo sequences
- increasing echo train length of FSE/TSE pulse sequences
- decreasing voxel size by increasing phase-encoding steps
- using STIR imaging to obtain fat suppression

Advanced pulse sequences for reducing metal artifact, including slice-encoding metal artifact correction (SEMAC) and multi-acquisition variable resonance imaging combination (MAVRIC), can be employed, although these measures may not be sufficient to correct the artifacts and thereby render unusable images. In these instances, MRI may not be the most suitable imaging modality.

Other Factors

In addition to the operator-controlled factors related to image quality, the scan operator is also responsible for ensuring the acquisition protocol maintains a safe level of RF and magnetic field exposure to the patient. Any approved clinical scanner estimates exposure prior to the scan and will not allow the scan to continue without adjustments. Still, it is important that the operator understands the parameters that can be manipulated so that imaging can be completed safely.

Field Strength Influence and Heating

While RF energy levels required during an MRI scan are quite low, the transmission of this energy into the body must be monitored to assure there are no adverse effects like heating of the tissue. This becomes more important at higher field strengths such as 3.0T, where the energy needed for an excitation pulse is four times greater than that for 1.5T.

For each protocol, the scanner estimates how much heating of tissues is expected by determining the specific absorption rate of the tissues to be scanned; refer to the glossary at the end of this material for specific absorption rate limits set by the FDA.

If the SAR exceeds a preset safety limit, the scanner automatically notifies the scan operator and suggests modifications to the protocol to reduce RF energy deposition. Generally, this can be accomplished by:

- selecting a longer TR to give the tissues time to cool between excitations
- reducing flip angles, which proportionately reduces power levels
- reducing the number of slices, which lowers the number of excitations needed for the full volume

dB/dt and Stimulation Limits

As gradient hardware has advanced and become faster, the rapid on-and-off switching of the gradients generates rapid changes in the magnetic field throughout the bore of the magnet. The rate of change of the magnetic field is measured as **dB/dt**, the change in magnetic field per unit time. As gradients are switched on and off more rapidly, the rate of magnetic field change increases. This is especially true when using pulse sequences for advanced applications such as diffusion-weighted imaging that rapidly switch large gradients.

A rapidly changing magnetic field induces electrical currents in a conductor, and a possible side effect in an MRI exam is that these currents may cause peripheral nerve stimulation. During the scan, the patient may experience sensations such as involuntary muscle contractions, flashing lights, or tingling in the fingertips or top of the head. As with SAR, scanners approved for clinical use monitor dB/dt levels and notify the scan operator if they exceed a specific threshold. The scan operator may choose to make protocol adjustments that reduce gradient amplitudes such as thicker slices or reduced diffusion gradients in order to proceed with the scan.

Fundamental Trade-offs in Image Quality and SNR

Voxel Size

Resolution in terms of the voxel size in the acquired image is a primary factor in the SNR of an image. The volume of material in each voxel is directly related to the number of hydrogen nuclei in that voxel. A voxel is three dimensional, whereas a pixel is a 2D element within a single image (**Figure 43**). Since the hydrogen nuclei create the signal in MRI, the more nuclei there are, the higher the signal. The voxel size can be simply calculated as the acquired pixel size in each of the two dimensions of the image (*x* and *y*, or frequency and phase) multiplied by the acquired slice thickness. An increase in pixel size in any direction, while keeping all other parameters the same, will increase SNR and image quality. Increasing FOV with the same matrix size also increases SNR.

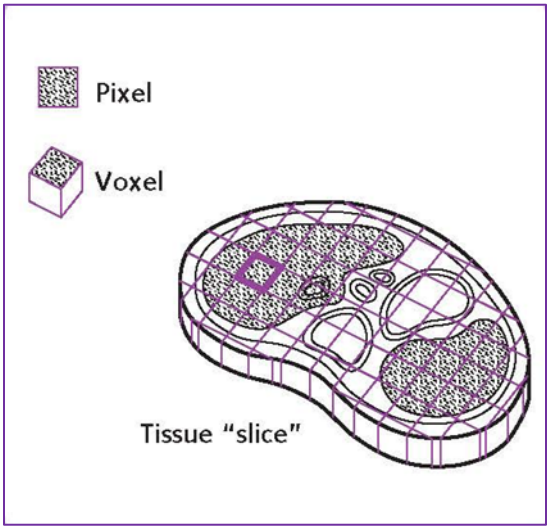

Figure 43. Each section of body tissue is imaged as a grid of cubes, or voxels. An MR image consists of light and dark pixels that correspond to tissue voxels.

Total Signal Recording Duration and Readout Time

The other major factor determining relative SNR of an acquired image is the total amount of time spent recording the signal during the scan. Note that this is not the same as the total scan time, as much of the scan time for one slice may be spent waiting for other slices to be acquired in interleaved scans or by applying other gradients and RF pulses to generate desired contrast. The time spent recording data is estimated by multiplying the number of phase-encode steps acquired, the number of averages, and the duration of each signal readout. This last parameter is not usually reported directly at the scanner, but it is proportional to one (1) divided by the bandwidth.

The SNR is proportional to the square root of the total readout time, meaning that if the total readout time is doubled (by doubling the number of phase encodes, the NEX/NSA, or by cutting the bandwidth in half), the SNR will increase by a factor of $\sqrt{2}$ or about 1.4.

Effect of smaller voxels

SNR is proportional to several common imaging parameters, shown in the following equation:

$$(x\ pixel\ size) \cdot (y\ pixel\ size) \cdot (slice\ thickness) \cdot \sqrt{\frac{(phase\ encodes)(NEX)}{(bandwith)}}$$

A smaller voxel size means fewer protons within each voxel and therefore less signal generation, which lowers the SNR. NEX/NSA will keep the signal the same but causes some of the noise to be canceled out, making the overall SNR higher. The lower the bandwidth, the more noise is filtered out of the signal while keeping the signal the same, which results in higher SNR. As with NEX/NSA, a greater number of phase encodes increases the duration of the signal recording, canceling out some of the noise.

It may be helpful to consider a few examples on how changing an image protocol affects image quality. Begin with an image with the following parameters:

- FOV 20cm × 20cm
- Matrix size 256 × 256, making a pixel size of about 0.8mm × 0.8mm
- Slice thickness 4mm
- 1 average (1 NEX)
- Bandwidth of 150 Hz/pixel

If we reduce the field of view to 16cm, this will decrease the pixel size by a factor of (16 ÷ 20) × (16 ÷ 20) = 0.64. The SNR will also drop by the same factor.

Effect of larger matrix and smaller voxels

If the matrix size is increased to 512 × 512, the pixel size shrinks to half in each direction, or about 0.4mm × 0.4mm. This creates a higher resolution image but at the cost of reduced SNR. However, with a matrix of 512, we will acquire twice as many phase-encode lines so the total readout time increases by a factor of two. The overall effect on SNR is a factor of 0.5 × 0.5 × √2 = 0.35.

Effect of increasing NEX/NSA

Increasing the number of averages from 1 to 2 increases the SNR by a factor of 1.4. Similarly, increasing the averages from 1 to 3 provides an increase of √3 = 1.7.

Another way to approach averaging is by determining the number of averages needed to create an image of higher resolution and the same SNR. For example, if slice thickness is reduced from 4mm to 2mm, the SNR will be reduced by a factor of 2. Four averages are needed to restore SNR back to the same level, since $\sqrt{4} = 2$.

Effect of higher bandwidth

Changes in bandwidth correspond to changes in the duration of signal readout after each excitation. Increasing the bandwidth is the same as decreasing the readout time and therefore will result in a decrease in SNR. If the bandwidth in our example were set to 300 Hz/pixel, the SNR would be decreased by a factor of $\sqrt{2} = 1.4$. Changes in bandwidth may require changes in TE or TR as well.

As always, there are multiple factors to consider when choosing MRI parameters.

Understanding Trade-Offs to Produce Consistent Quality

It is essential that scanner operators not only have a full understanding of the factors affecting SNR but how changing scan parameters affects image quality. When parameters such as resolution or bandwidth are changed, other parameters may need to be adjusted to ensure high-quality images (**Tables 4 and 5**). On some systems, the scanner may report how the SNR is affected as parameters are changed from an initial protocol to ensure that image quality is kept sufficiently high.

Changing Scan Parameter		Leads To		Results In
Increase pixel/voxel size *(no change in matrix)*	▶	Larger FOV	▶	Increased SNR Decreased resolution
Increase matrix *(no change in FOV)*	▶	Smaller voxels	▶	Decreased SNR Increased resolution
Decrease voxel size *(no change in FOV)*	▶	Larger matrix	▶	Decreased SNR Increased resolution
Increase NEX	▶	Longer scan time	▶	Increased SNR Same resolution
Increase bandwidth	▶	Shorter readout or echo time	▶	Decreased SNR Same resolution
Reduce slice thickness	▶	Smaller voxels Same pixel size	▶	Decreased SNR Increased resolution

Table 4. Trade-offs in changing MR imaging parameters.

Matrix Size (read x phase)	Field of View (read x phase) cm x cm	Slice Thickness mm	Averages	Bandwidth Hz/pixel	Relative SNR
256 x 256	24 x 24	4	1	150	100
256 x 256	16 x 16	4	1	150	44
256 x 256	24 x 24	3	1	150	75
256 x 256	24 x 24	4	2	150	141
256 x 256	24 x 24	4	1	300	71
512 x 256	24 x 24	4	1	150	50
512 x 512	24 x 24	4	1	150	35

Table 5. Effect of common settings on SNR.

SUMMARY

We have described many of the imaging parameters that all MRI technologists should fully comprehend in order to accurately and efficiently complete a high-quality, diagnostic MR examination. Understanding the many factors that influence the appearance and quality of MR images may seem daunting, but it is this versatility that makes MRI a uniquely rich imaging tool.

Several of these factors are under the control of the scan operator, such as slice location and dimension, resolution, and timing parameters. In many cases, the initial selection of these parameters will be set by established imaging protocols.

Understanding the many factors that can influence the appearance and quality of MR images may seem daunting, but it is this versatility that makes MRI a uniquely rich imaging tool.

There is also a range of conditions that causes poor image quality and artifact in MR images that may stem from hardware and installation problems or from patient-specific issues.

It is vital that imaging technologists understand how to adapt protocols to ensure a high-quality study for the variety of patients and conditions seen in practice. Similarly, while other factors such as artifacts and SAR limits cannot be controlled, their effects can be mitigated.

Consistently achieving the best possible diagnostic image quality, balanced with the shortest possible scan time and clinical diagnostic need, is one of the most challenging and rewarding aspects of the MRI technologist experience. Translating your knowledge into practice is not only essential for providing a safe and comfortable patient exam but for becoming a true working partner with the radiologist to produce the diagnostic images that will guide patient management and treatment.

POINTS for PRACTICE

1. What are the three general characteristics that describe the quality of an MR image?

- **Spatial resolution** defines how much detail is captured in an image and depends on matrix size and field of view.
- **Signal-to-noise ratio (SNR)** is the proportion of RF signal actually used to construct an image relative to the amount of random background noise. A high-quality image has a high SNR; a low-quality image has a low SNR and appears grainy.
- **Image contrast** is the difference in the brightness between two structures or two regions within an image.

Changes in one of these parameters affect one or both of the others, meaning that optimizing MR image quality of one parameter may have negative effects in one or both of the others.

2. What are some parameters that affect image quality that are not scan operator-controlled?

- field homogeneity
- gradient strength and speed
- field strength
- physical characteristics of the scanned tissues
- flow and motion

3. Name some parameters that can affect image quality that are scan operator-controlled.

- bandwidth
- field of view
- flip angle
- interslice gap
- number of averages
- patient positioning
- resolution and matrix size
- scan time
- selection and positioning of RF coils
- slice location and orientation
- slice thickness

4. Provide some examples of artifact.

- aliasing
- chemical shift
- ghosting
- motion
- slice overlap (crosstalk)
- susceptibility

5. As the scan operator, you must decide what variables are essential for obtaining the best images. What are some of the trade-offs needed to produce consistent quality?

- bandwidth
- field of view
- matrix size
- NEX
- slice thickness
- TE
- TR
- voxel size

6. Increasing matrix size and decreasing FOV both alter image quality criteria in the same way. What are the criteria? What accounts for the change?
Spatial resolution increases with decreased FOV and/or increased matrix size. Spatial resolution depends on the number of pixels representing a specific area; a greater number of pixels produce greater resolution. By decreasing FOV, which decreases the area of body tissues to be imaged, more pixels are made available to depict the remaining areas of tissue; by increasing matrix size, more pixels are used to generate the image. Both of these changes reduce the SNR.

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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
FIESTA	fast imaging employing steady-state acquisition (GE)
FISP	fast imaging with steady-state precession (Siemens)
fMRI	functional MRI
FLAIR	fluid-attenuated inversion recovery
FOV	field of view
FS	fat suppressed (also FatSat)
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
NEX	number of excitations (also NSA)
NMR	nuclear magnetic resonance
NSA	number of signal averages (also NEX)
NSF	nephrogenic systemic fibrosis
PC	phase contrast
PDW	proton density-weighted
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 (eg, FISP, FIESTA, balanced FFE)
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
TE/2	echo time halved
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 selected field of view 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 area of interest 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 which oscillates at the Larmor frequency

bandwidth
refers to a range of frequencies; this range helps determine slice thickness and signal recording time

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 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 both generate magnetic fields (eg, RF transmit coils and gradient coils) and detect changing fields (receiver coils)

contrast agent
a substance introduced into the body to increase the contrast of tissues or fluids, especially blood vessels. MRI contrast agents are usually gadolinium-based. When injected into the blood stream, the tissues into which the agent penetrates will appear bright on T1-weighted imaging.

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, nitrogen, copper, gold

diffusion tensor imaging (DTI)
MRI technique that maps the 3D diffusion of water molecules throughout the brain

diffusion-weighted imaging (DWI)
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 factor

electromagnetic spectrum
continuous range 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 Richard R. Ernst (1933-), a Swiss physical chemist 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 used to generate a measurable signal after excitation 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 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, eg, 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)
an 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; 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 which highlights active structures during particular mental operations

gadolinium-based contrast agent (GBCA)
gadolinium (Gd) is a rare earth element that when chelated is used as a paramagnetic contrast agent in MRI; administration of a GBCA will shorten T1, resulting in tissues that take up the agent appearing brighter on T1-weighted imaging

ghosting
an image artifact where a shifted copy of the object or “ghost” appears elsewhere in the image; commonly caused by 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 variable strength depending on the location within the field. In MRI the gradient field is generated by gradient coils. Gradient fields add to or subtract from the main magnetic field and are used for slice selection, frequency encoding, and phase encoding, altering the MR signal depending on location within the magnet. Also referred to as “gradient.”

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 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

image contrast
the difference in brightness between two structures or regions within an image; one of the three primary requirements of a highly diagnostic MRI exam: spatial resolution, SNR, and image contrast

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

interleaving
arranging the order of 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; often used to suppress the signal from a particular tissue such as fat or CSF based on its T1 relaxation time. See 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

isocenter
the center of the magnet bore and main magnetic field where the field and gradients are the most homogeneous. The area of interest should be positioned close to isocenter if possible for the best image quality

***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 ($f = \gamma B_0$)
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; 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

magnets, types of
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.
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.
superconducting – most commonly used in MR scanners. Superconducting magnets 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

nuclear magnetic resonance (NMR)
a physical phenomenon in which nuclei in a magnetic field absorb and re-emit electromagnetic radiation

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 of the imaging plane that is not a multiple of 90° as opposed to an orthogonal angle, which is a right-angle. In MRI, a *single oblique* angle is an angle that is not 90° to one of the *x, y,* or *z* axes. A *double oblique* angle is when two of the angles are not 90° to the third axis.

oversampling
increasing the sampling FOV for 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 coil elements to accelerate the acquisition of images to reduce 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 both 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 image contrast is not dependent on T1 or T2 but is dependent on the number of hydrogen protons that are 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 describe 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-like 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/time to recovery (TR)
time interval between two RF excitation pulses in an MRI pulse sequence

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 excitation 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.

shimming
technique used to eliminate inhomogeneity in the main magnetic field; performed prior to certain exams by measuring and adjusting the main magnetic field with additional coils installed in the scanner (active shimming). Passive shimming uses small pieces of metal to adjust the magnetic field when the magnet is installed.

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 called 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; 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)
pulse sequence most closely related to gradient-echo imaging as there is no refocusing excitation pulse. 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 time or echo delay time; time interval between the initial RF excitation pulse and the echo of a spin-echo or gradient-echo pulse sequence

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 affect the amount of longitudinal magnetization; also time to inversion

time-of-flight imaging (TOF)
a pulse sequence that makes stationary materials appear dark on the image, while moving tissues such as blood appear 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

tractography
fiber tracking of the brain's white matter tracts using diffusion tensor imaging

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 in a single repetition. 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