MRI for Technologists

MRI Systems and Coil Technology

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.

Original Release Date: **October 2006**
Review Date **June 2018**
Expiration Date: **November 1, 2019**

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.

MRI for Technologists: MRI Systems and Coil Technology introduces the learner to different bore types and field strengths, types of MR coils, and appropriate coil selection for optimizing the MR examination.

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

- State the advantages of the short bore MR system
- Compare advantages and disadvantages of the low-field and high-field strength magnets
- Decide what type of magnet is most suitable for a specific MR examination
- Describe the major types of MR coils and when they should be used
- Discuss advantages and disadvantages of specific MR coils
- Select appropriate coil configuration to optimize the MR examination

EDUCATIONAL CREDIT

This program has been approved by the American Society of Radiologic Technologists (ASRT) for 1.0 hour ARRT Category A continuing education credit.

HOW TO RECEIVE CREDIT

Estimated time to complete this activity is **1.0** hour. 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 www.icpme.us.
- Read the entire activity.
- Log in to your account at www.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 certificate of credit from your online account.

ACKNOWLEDGMENTS

Our appreciation goes to Thomas Schrack, BS, ARMRIT for his review and update of this material. We would also like to acknowledge the original authors of this material:

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FDA Drug Safety Communication: FDA warns that gadolinium-based contrast agents (GBCAs) are retained in the body; requires new class warnings

https://www.fda.gov/Drugs/DrugSafety/ucm589213.htm Accessed June 14, 2018.

05-16-2018 Update

In addition to approving the updated prescribing information concerning the gadolinium retention safety issues described in the Drug Safety Communication below, FDA has also approved new patient Medication Guides for all GBCAs.

Health care professionals and patients can access the patient Medication Guides according to the GBCA drug name* on the Medication Guides webpage, or the latest prescribing information by searching in Drugs@FDA.

All MRI centers should provide a Medication Guide the first time an outpatient receives a GBCA injection or when the information is substantially changed. In general, hospital inpatients are not required to receive a Medication Guide unless the patient or caregiver requests it. A health care professional who determines that it is not in a patient's best interest to receive a Medication Guide because of significant concerns about its effects may direct that it not be provided to that patient; however, the Medication Guide should be provided to any patient who requests the information.†

*The brand names of the GBCAs can be found in Table 1 below.
†For more information on distribution of Medication Guides, see the Guidance Document, the Drug Info Rounds Video, or the Code of Federal Regulations at 21 CFR 208.26.

This is an update to the FDA Drug Safety Communication: FDA identifies no harmful effects to date with brain retention of gadolinium-based contrast agents for MRIs; review to continue issued on May 22, 2017.

12-19-2017 Safety Announcement

The U.S. Food and Drug Administration (FDA) is requiring a new class warning and other safety measures for all gadolinium-based contrast agents (GBCAs) for magnetic resonance imaging (MRI) concerning gadolinium remaining in patients' bodies, including the brain, for months to years after receiving these drugs. Gadolinium retention has not been directly linked to adverse health effects in patients with normal kidney function, and we have concluded that the benefit of all approved GBCAs continues to outweigh any potential risks.

However, after additional review and consultation with the Medical Imaging Drugs Advisory Committee, we are requiring several actions to alert health care professionals and patients about gadolinium retention after an MRI using a GBCA, and actions that can help minimize problems. These include requiring a new patient Medication Guide*, providing educational information that every patient will be asked to read before receiving a GBCA. We are also requiring manufacturers of GBCAs to conduct human and animal studies to further assess the safety of these contrast agents.

GBCAs are used with medical imaging devices called MRI scanners to examine the body for problems such as cancer, infections, or bleeding. GBCAs contain gadolinium, a heavy metal. These contrast agents are injected into a vein to improve visualization of internal organs, blood vessels, and tissues during an MRI, which helps health care professionals diagnose medical conditions. After being administered, GBCAs are mostly eliminated from the body through the kidneys. However, trace amounts of gadolinium may stay in the body long-term. Many GBCAs have been on the market for more than a decade.

Health care professionals should consider the retention characteristics of each agent when choosing a GBCA for patients who may be at higher risk for gadolinium retention (see Table 1 listing GBCAs). These patients include those requiring multiple lifetime doses, pregnant women, children, and patients with

inflammatory conditions. Minimize repeated GBCA imaging studies when possible, particularly closely spaced MRI studies. However, do not avoid or defer necessary GBCA MRI scans.

Patients, parents, and caregivers should carefully read the new patient Medication Guide* that will be given to you before receiving a GBCA. The Medication Guide explains the risks associated with GBCAs. Also tell your health care professional about all your medical conditions, including:

- If you are pregnant or think you might be pregnant
- The date of your last MRI with gadolinium and if you have had repeat scans with gadolinium
- If you have kidney problems

There are two types of GBCAs based on their chemical structures: linear and macrocyclic (see Table 1 below). Linear GBCAs result in more retention and retention for a longer time than macrocyclic GBCAs. Gadolinium levels remaining in the body are higher after administration of Omniscan (gadodiamide) or OptiMARK (gadoversetamide) than after Eovist (gadoxetate disodium), Magnevist (gadopentetate dimeglumine), or MultiHance (gadobenate dimeglumine). Gadolinium levels in the body are lowest after administration of Dotarem (gadoterate meglumine), Gadavist (gadobutrol), and ProHance (gadoteridol); the gadolinium levels are also similar across these agents.

*The Medication Guide will be posted once it is approved.

Table 1. FDA-Approved GBCAs*

Brand name	Generic name	Chemical Structure
Dotarem†	gadoterate meglumine	Macrocyclic
Eovist	gadoxetate disodium	Linear
Gadavist†	gadobutrol	Macrocyclic
Magnevist	gadopentetate dimeglumine	Linear
MultiHance	gadobenate dimeglumine	Linear
Omniscan‡	gadodiamide	Linear
OptiMARK‡	gadoversetamide	Linear
ProHance†	gadoteridol	Macrocyclic

*Linear GBCAs result in more gadolinium retention in the body than macrocyclic GBCAs.
†Gadolinium levels remaining in the body are LOWEST and similar after use of these agents.
‡Gadolinium levels remaining in the body are HIGHEST after use of these agents.

To date, the only known adverse health effect related to gadolinium retention is a rare condition called nephrogenic systemic fibrosis (NSF) that occurs in a small subgroup of patients with pre-existing kidney failure. We have also received reports of adverse events involving multiple organ systems in patients with normal kidney function. A causal association between these adverse events and gadolinium retention could not be established.

We are continuing to assess the health effects of gadolinium retention in the body and will update the public when new information becomes available. We are requiring the following specific changes to the labeling of all GBCAs:

- A *Warning and Precaution*
- Changes related to gadolinium retention in the *Adverse Reactions, Pregnancy, Clinical Pharmacology*, and *Patient Instructions* sections

We urge patients and health care professionals to report side effects involving GBCAs or other medicines to the FDA MedWatch program.

MRI Systems and Coil Technology

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MRI SYSTEMS

Three primary types of magnets are used in MRI systems: permanent, resistive, and superconducting. The operating principles and construction details of each type of magnet are detailed in other materials in this series. In this unit we will discuss the particular abilities and disadvantages of each magnet when performing different types of MR examinations.

In addition to the primary type of magnet, clinical MR scanners can be broadly divided into two categories: low-field strength and high-field strength. Scanners having a magnetic field strength below 1.0 T are considered low-field, while scanners with field strength of 1.0 T or greater are high-field systems.

FIELD STRENGTH

Low-field Scanners

Low-field scanners can be built around any of the three primary magnet types. Superconducting magnets are used for field strengths at the upper end of the low-field category (approximately 0.5 T and above) will be discussed in the high-field scanner section.

Permanent Magnet Systems

In a permanent magnet system, the magnetic field is generated vertically between a pair of plates, and the patient is positioned horizontally between those plates. Commercial permanent magnet systems are capable of generating a magnetic field of up to approximately 0.3 T. The magnetic field in a permanent magnet system is always in effect and

requires no power or cooling to maintain. The magnetic field is stable, provided that its temperature is precisely controlled. However, permanent magnets have poorer field homogeneity compared to superconducting magnets, and the volume of homogeneous field is also somewhat smaller. The weight of a permanent magnet is also enormous at 15,000– 20,000 kg and may require reinforced flooring to site the system, particularly if the magnet is not located at ground level.

Resistive Magnet Systems

The magnetic field in a resistive magnet is generated with an electromagnet. The electromagnet requires a large electric current (approximately 50 kilowatts for a 0.15 T field) to be passed through electrical windings to produce the magnetic field. As a result, large amounts of heat are generated, so the scanner must have a substantial cooling system. Like permanent magnet systems, the homogeneity of the field and the overall volume of homogeneous field of a resistive magnet are reduced compared to a superconducting magnet. Resistive magnets are available in field strengths to about 0.3 T. While a resistive magnetic field can be turned off when not in use, the cost of electricity when in use (in addition to expensive cooling systems) can be quite high. Moreover, when the system is first turned on (meaning electrical current is turned on after being off for some amount of time), it requires as much as 30 to 60 minutes to bring up the magnetic field and then to stabilize it, which causes a loss in productivity and efficiency.

Low-field scanners are used mostly in systems with an open bore design, in systems designed to simultaneously perform interventional procedures, and for small specialized systems such as extremity MRI systems.

Superconducting Magnet Systems

High-field scanners are always built around superconducting magnets, and the majority of installed MRI systems use this type of magnet. Clinical scanners with superconducting magnets are commercially available with field strengths up to 3.0 T, and whole-body systems currently exist at research facilities with fields as high as 9.4 T.

Of the three magnet types, superconducting magnets are capable of generating the most homogeneous magnetic field over the largest volume. The magnet orientation is nearly always a traditional cylindrical bore configuration, although the length of the bore may vary considerably. Superconducting magnets with open bore design have appeared in recent years as well, usually with field strengths between 0.5 T and 1.0 T (Figures 1 and 2).

A superconducting magnet is an electromagnet where the windings are made of a superconducting material. Materials become "superconductive" when cooled to temperatures approaching 0° Kelvin. At this temperature all molecular motion stops. Since the

Figure 1. Example of an open bore permanent magnet system.

a b c

Figure 2a-c. Examples of high-field 1.5 T MRI systems.

preponderance of any compound is actually space created by the electron cloud, any electricity passed through such a material will encounter no resistance. The material used in superconducting MRI systems becomes superconductive at 4° Kelvin when immersed in liquid helium. As long as this temperature is maintained, the magnetic field will also persist without any additional electrical power. As a result, the magnetic field is always present, without the need for continued electric voltage. In essence, the electricity continues to flow indefinitely, even without a primary source. In an emergency, the field can be turned off by slowly discharging the current flowing through the windings, which may require 30 minutes or more. This is known as "ramping down" the magnet. An immediate release of the cryogens—whether on purpose in response to a life-threatening emergency or by accidental induction of excessive heat into the magnet—is known as a "quench." An uncontrolled quench can be powerful and dangerous because super-cooled liquid helium expands exponentially and displaces breathable oxygen from the room. If the cryogen ventilation system is overpowered by the sheer amount of escaping helium, the oxygen in the room can drop to levels below that needed to sustain human life.

The structure that contains the magnet and the cryogens is called the cryostat. The cryostat is insulated from the surrounding environment as much as possible to reduce heating of the cryogens. However, the liquid helium will slowly boil off and eventually requires refilling. In modern systems, a helium refill may not be required for as much as a year or more.

FIELD STRENGTH CONSIDERATIONS AND COMPARISONS

The magnetic field strength of a scanner affects a wide variety of imaging factors that must be taken into account when considering the most appropriate system and protocol to use for any specific examination (Table 1).

Signal-to-Noise Ratio

The signal-to-noise ratio (SNR) is significantly greater at high fields as compared to low fields. In general, the signal-to-noise ratio increases in direct proportion to the main magnetic field strength. For example, with all other factors remaining equal, the SNR of

Parameter	As Field Strength Increases	Notes
Acoustic Noise	Increases	Higher field exerts greater force on conductors
Fat/Water Separation	Increases	Easier to perform fat saturation, but increases chemical shift artifact
SAR	Increases	4X increase for 2X increase in field
SNR	Increases	Directly proportional to field strength
Susceptibility	Increases	Yields more artifacts in fast scanning but improves *f*MRI
T1	Increases	Requires longer TR or reduced flip angle for same contrast

Table 1. **A summary of the effects of different field strength systems.**

an exam performed at 3.0 T will be twice that of the same exam at 1.5 T. However, other hardware considerations such as bore size and coil design may also affect SNR between systems as well.

The increased SNR available at high fields for equivalent examinations may also permit trade-offs in the parameters for the exam. Image SNR is affected by the resolution, signal bandwidth, and other parameters. The higher SNR at high-field can be "traded" for improvements in one or more of these factors. For example, at 3.0 T it should be possible to acquire images with the same SNR and examination time, but with higher resolution than at 1.5 T. Alternatively, the same SNR and resolution of the 1.5 T exam could be acquired in a shorter amount of time. Coils and other configuration details may affect the degree to which this is possible.

Specific Absorption Rate

A limiting factor of high-field imaging, especially at 3.0 T, is the increased energy disposition for the same types of RF pulses. For an increase of a factor of two in field strength from 1.5 T to 3.0 T, RF energy deposition increases by a factor of four. Simply put, the higher the field strength, the more RF energy is needed to flip a proton away from a longitudinal orientation. This is especially troublesome for pulse sequences that use many 180° pulses, such as fast spin-echo (FSE). Restrictions on SAR may reduce the number of slices or increase the repetition time (TR) required (increasing the scan time) to achieve the desired coverage.

Modifications to the acquisition protocol that can reduce SAR for high-field exams include:

- increased RF pulse spacing
- decreased flip angles
- use of hyperechoes
- use of parallel imaging

Field Homogeneity

Imaging at higher field strengths such as 3.0 T can introduce greater magnetic field inhomogeneities compared to 1.5 T imaging. Another factor is the varying magnetic susceptibility in the human body itself, which has a greater influence at higher field strengths. Inhomo-

geneities in the main magnetic field can result in image distortions, blurring, loss of resolution, and a reduction in the usable field of view (FOV) in the magnet. Inhomogeneity can also adversely affect fat saturation due to the corresponding chemical shifts. High quality shimming is a necessity to minimize the effects of inhomogeneity at higher fields.

Magnetic Susceptibility Effects

Magnetic susceptibility is the ability of a tissue to magnetize. Some tissues magnetize to a great degree (blood, fat), while other tissues (compact bone) do not. Magnetic susceptibility effects can be either desirable or undesirable. The effect of susceptibility can create needed contrast in tissues or cause artifacts that are detrimental to the image quality.

Susceptibility effects are increased as imaging moves to higher field strengths. This can be either a disadvantage or an advantage, depending on the specific imaging application.

In applications with areas of high susceptibility—such as the air-tissue interfaces around the sinuses and near the heart-lung boundary in cardiac imaging—many acquisitions, especially fast imaging acquisitions like echo-planar and spiral imaging, may suffer greater distortions and signal loss compared to imaging at lower field strengths.

Some techniques to reduce susceptibility artifacts include:

- reduction in echo time
- reduction in echo spacing
- increase in bandwidth
- use of parallel imaging

Pulse sequences that rely on susceptibility effects to generate image contrast, however, benefit from higher field imaging. Blood Oxygen Level–Dependent (BOLD) imaging contrast results from the susceptibility effects of oxygenated blood, which are increased at higher field strengths. Perfusion studies using dynamic susceptibility contrast (DSC-MRI) may also benefit from the greater susceptibility effects. MR spectroscopy is well known to improve greatly at higher field strengths. There are two main reasons: higher SNR allows signals of weaker metabolites to be visualized, and higher field strength equates to greater chemical shift between metabolites, yielding more distinctive spectroscopy peaks from neighboring metabolites.

Other Imaging Effects

The main magnetic field strength affects several tissue parameters. T1 increases with higher field strength. T2 is generally reduced, though the magnitude of the change is highly variable. T2* (T2 star) is also reduced, primarily due to inhomogeneity and susceptibility effects. The chemical shift between fat and water grows proportionately with field strength.

These effects, as well as many of the other changes that depend on field strength mentioned above require that imaging protocols be designed for the specific field strength to be applied. Protocols that yield acceptable images at one particular field strength may give poor results or may not be possible at a different field strength. Image contrast can change considerably if appropriate changes are not made in the image acquisition parameters.

Coil Selection and Availability

Coils for specialized applications must be designed for a specific MR system model and field strength. Since 1.5 T systems continue to be the most commonly used, coil availability and quality is greatest for these systems. Lower field strength scanners are normally

used for routine examinations. In the case of special purpose systems, such as extremity scanners, coil selection is not a large issue because the scanner is designed for only a narrow range of applications.

3.0 T scanning has become widely clinically available in the last few years. Coil selection for 3.0 T scanners is presently more limited than that for 1.5 T. Initial applications of clinical 3.0 T imaging have been mostly in the head, so head coils were the first to be developed. Other areas, such as musculoskeletal imaging, lagged in coil development. The trend towards more 3.0 T clinical sites has spurred the development of newer high-channel coils. Now that coil development for these applications is occurring, applications such as body and MSK imaging are greatly benefiting from 3.0 T imaging. As a result, 3.0 T imaging is now a mainstay in many imaging facilities.

Acoustic Noise

The source of acoustic noise during an MR examination is the interaction of the large static magnetic field with the gradient coils. The pulsing of current through the gradient coils in the main magnetic field causes the coils to vibrate, much like a loudspeaker. The volume of the noise depends on both the construction and mounting of the gradient coils, as well as the gradient amplitudes and timing for any particular pulse sequence. Increasing the magnetic field will also increase the amplitude of these forces, and thus the noise. Patient use of earplugs or headphones is vital to prevent hearing damage, and usage becomes more important at higher field strengths and during examinations using higher performance gradients. Soundproofing the magnet room is also important for MR personnel in the surrounding environment.

SUMMARY OF FIELD EFFECTS

It is important to understand the physical limits of lower field strength systems. All MR imaging requires the technologist to balance the elements of spatial resolution, SNR, and scan time, although what is considered an acceptable balance is somewhat subjective. Higher field strength systems have intermittently higher SNR. Low-field strength scanners have an important place in medical imaging; however, it must be remembered that to achieve the same diagnostic quality as a high-field system using a low-field system (all things being equal), at least one of the following, or some combination thereof, must occur:

- the scan time must be increased
- the images will have lower SNR and/or resolution
- fewer images will be obtained

SCANNER BORE DESIGN

Scanners can also be categorized by the type of bore for patient entry into the system. The two main types are cylindrical bore and open bore systems.

Cylindrical Bore

Virtually all high-field systems (1.5 T and higher) are cylindrical bore systems. A cylindrical bore magnet has a tube-shaped opening in the main magnet. The patient is placed on a motorized table that slides the region of interest (ROI) into position in the center of the tube, so that the patient is surrounded on all sides. The length and diameter of the bore vary widely among currently used MRI systems.

Older high-field MRI scanners (typically produced before 1995) often have a longer bore. The bore in these systems may be as long as 250 cm, which means the patient will be completely inside the bore for almost all types of examinations. The length was required to generate a field in the center of the bore homogeneous enough for high quality imaging. However, the longer bore magnets can be problematic for even mildly claustrophobic patients.

In recent years a great deal of effort has gone into magnet design to shorten the length of the bore. The theory is that the shorter bore will result in the lower likelihood of claustrophobia for patients. However, to date there have been no studies that either prove or disprove this theory.

Newer systems have a much shorter bore length, typically in the range of 150–180 cm; a 1.5 T magnet with a bore length of just 125 cm was recently released. The smaller overall size of the machine makes it easier to install, because less space is required for siting. The homogeneous region of the field remains about the same as in older long bore systems; however, the shorter bore magnet designs have a more difficult time maintaining high homogeneity over a large FOV. As a result, for some shorter bore systems the maximum useable FOV is reduced as compared to older, long bore systems.

Another important variable in cylindrical bore systems is the bore diameter. The diameter reflects the size of the opening for the patient; typically, this is in the range of 55–60 cm, though sizes as large as 70 cm have recently become available for clinical use. The actual FOV that can be achieved for imaging is somewhat smaller, around 40–50 cm.

Open Bore

An open bore magnet design uses a pair of magnet coils arranged vertically, with a relatively large gap in the center for patient access. The patient slides in from the side and is not completely surrounded by the system. The open bore accommodates larger subjects and lessens the feelings of claustrophobia in some patients. Patients who could not be scanned in a cylindrical bore system due to size or claustrophobia may be able to tolerate an open bore magnet.

Open bore systems can use any of the three major magnet types, although permanent magnets are most common. Open bore systems are most often low-field, which makes them easier to install and popular in outpatient settings due to the relatively small size of the system. Due to the limitations of low-field systems, however, open bore scanners are not always an option for patients who might prefer them for reasons of comfort.

Considerations

As a rule, the higher field strength and better field homogeneity of cylindrical bore magnets results in considerably higher image quality when compared to open bore magnets. The two main advantages of open bore systems are for obese patients and for those who experience claustrophobia. The design and "feel" of open bore systems may permit many patients to be scanned that might otherwise need sedation or be unable to tolerate the scan. Open bore designs can accommodate much larger patients than some closed bore systems. Low-field systems also offer advantages such as less visible motion artifacts, less SAR to the patient (allowing more slices per TR), and a far lower acoustic noise level for the patient. However, the lower field, longer examination times, and limited range of imag-

ing protocols mean that open bore systems may not be appropriate for a particular examination. The continuing trend of cylindrical bore systems with shorter and wider bore sizes (comparable to a CT scanner) may eventually permit high-field scanning of both large and potentially claustrophobic subjects.

GRADIENTS

The gradient system of an MR scanner generates an additional set of magnetic fields that vary depending on location within the magnet. The dependence of these fields on location is used to determine the spatial variation in the signals given off by the tissues. Therefore, it is the gradient fields that make it possible for the system to create 2D and 3D images of tissues.

The gradient system of a scanner is made up of three sets of coils. Each coil can generate a magnetic field that varies along one of the three principal directions of X, Y, and Z (Figure 3). The change in the magnetic field produced by the gradient is directly proportional to the distance from the center of the magnet and the strength of the gradient applied. The gradients are switched on and off at a rapid rate throughout the scan, and the amplitude of the fields they generate can be varied to acquire enough data to generate a 2D image or 3D volume. Several characteristics of the gradient system impact their performance and the types of pulse sequences that can be applied.

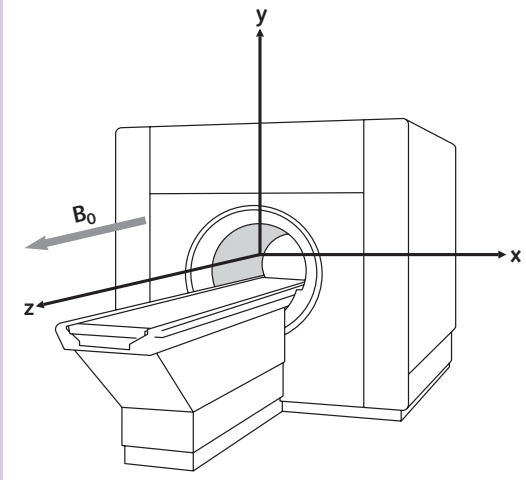


Figure 3. The three primary directions in the cylindrical, or traditional, magnet.

Amplitude

The gradient amplitude measures the maximum variation in the magnetic field that the gradient coils are capable of creating. In the past this was referred to as gradient strength. However, because other gradient attributes also describe gradient performance, eg, rise time and slew rate, employing the term "strength" to characterize gradient amplitude is not very useful. The maximum amplitude is measured in units of mT/m, with typical values ranging from 10–60 mT/m in high-field clinical scanners; however, the gradient coils can be programmed to create any amplitude up to the maximum value. For example, a 10 mT/m gradient amplitude using the X gradient (the X direction is the left-right direction looking into the scanner) means that the magnetic field will be 1.0 mT higher at a location 10 cm to the right of the center of the magnet (10 mT/m times 0.1 meter).

The amplitude of the gradient affects the spatial resolution and FOV that the system is capable of creating. A higher maximum gradient will allow the system to acquire images using thinner slices. Higher gradient amplitudes are also needed to perform rapid imaging sequences such as echo-planar imaging (EPI).

Slew Rate

The gradient fields are not turned on constantly but pulsed on and off during image acquisition to obtain the needed data. Switching from the off condition to the maximum gradient cannot happen instantaneously; it takes some time. The time required to switch the gradients from zero amplitude to the maximum can be measured two ways: rise time and slew rate.

The rise time is defined as the time required to switch the gradients from zero to their maximum amplitude. A typical rise time is between 0.1 to 1.0 ms on many commercial systems. This time lag is particularly important for fast sequences such as EPI, where the gradients are switched back and forth many times during a single data acquisition. The faster the rise time, the more rapidly the image can be acquired for these sequences. In EPI, where the gradients go from max positive to max negative, the switching of the gradient from –max (max –G) to + max (max +G) is referred to as ramp time (Figure 4).

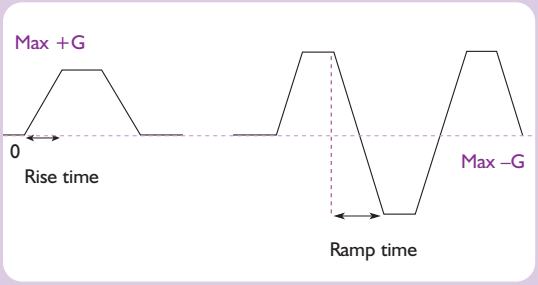


Figure 4. Gradient *rise time* and gradient *ramp time* should not be confused. Rise time is the time required for a gradient to go from zero to its maximum amplitude (max G). Conversely, ramp time is the time required for a gradient to go from its maximum positive amplitude (max +G) to its maximum negative amplitude (max –G). Since EPI sequences rely heavily on gradient reversals in the echo train, ramp time becomes far more relevant than rise time.

Another way of describing gradient performance is the slew rate of the gradients, which measures the maximum possible change in the gradient per unit time. The units of this measurement are then either milliTesla per meter per millisecond (mT/m/ms) or Tesla per meter per second (T/m/s); these units will actually have the same value. For example, a gradient system with 40 mT/m maximum amplitude and a 0.4 ms rise time would have a maximum slew rate of 100 mT/m/ms or 100 T/m/s (40 mT/m divided by 0.4 ms, or 0.04 T/m divided by 0.0004).

The slew rate will also determine how fast the magnetic field changes while the gradients are switching on and off. For some pulse sequences, the slew rate that can actually be used may be restricted by FDA regulations. This restriction is designed to limit peripheral nerve stimulation, not the capabilities of the gradient coils and amplifiers.

Duty Cycle

To rapidly switch the gradients on and off requires large amounts of power to be provided to the gradient amplifiers, resulting in the generation of substantial amounts of heat.

The duty cycle of the gradients specifies the fraction of time that the gradients can be turned on compared to the total scanning time. A duty cycle of 100% means that the gradients can be switched on and off continuously. A duty cycle of less than 100% means that the gradients must "rest" for some time during the pulse sequence in order to prevent the accumulation of too much heat, which could damage the system. The duty cycle may determine the shortest TR that is possible with

a particular pulse sequence to make sure that the gradients are not switched on too often. All major manufacturers of clinical MRI systems currently report 100% duty cycles on their systems.

IMAGING CAPABILITIES

As noted above, the capabilities of the gradient system will partly determine the imaging parameters that can be achieved. The maximum gradient amplitude directly affects the minimum slice thickness that can be prescribed. Along with the RF system, it may also influence the highest resolution the system is capable of in a given scan time. The slew rate will also have an effect on how fast images can be acquired. In most sequences, no image data are acquired while the gradients are ramping up to their maximum or back down to zero, so a high slew rate minimizes this dead time in the acquisition and may permit a shorter repetition to be achieved. The duty cycle allowed sets limits on the TR and acts as another factor in determining the rate at which images can be acquired.

The primary applications that demand high-performance gradients are fast scanning techniques such as EPI, fast 3D gradient-echo imaging, and extremely rapid FSE imaging. In EPI, the amount of data acquired after each excitation pulse is much greater than for a standard spin-echo, or even for a gradient-echo pulse sequence. EPI is achieved by very rapidly switching the gradients from positive to negative amplitudes to acquire multiple lines of data from the same excitation. In the extreme case, entire images can be acquired in a single excitation, which requires the system to switch the gradients from positive to negative 128 or more times within 50–200 ms. Such fast acquisition requires large amplitude gradients to acquire data with high bandwidth, high slew rates to minimize the amount of time between the positive and negative gradient acquisitions, and a high duty cycle to repeat the process many times in a short repetition time.

SITING CONSIDERATIONS

The strong magnetic field associated with the MR scanner requires that care be taken in the siting of the magnet. It is important to consider the location and the surrounding environment, both to assure that interference and artifacts do not arise as a result of outside conditions, as well as to guarantee that the magnetic field neither poses a safety hazard nor affects other nearby instrumentation.

Magnetic Shielding

Magnetic shielding refers to the methods used to keep the strong magnetic field confined to a small area surrounding the magnet, preferably within the room where the magnet is housed. The magnetic field surrounding the magnet is known as the fringe field. The fringe field is primarily an issue of safety, as the magnetic field can attract objects from a distance, causing them to fly towards the magnet. The fringe field can also cause damage to pacemakers and other implanted devices. In practice, the safe area around a magnet is defined as the region outside which the magnetic field is less than 5 Gauss (0.5 mT); this is considered the safety limit for unrestricted exposure to regular public traffic. In general, higher field strength magnets will have a larger fringe field.

There are two general categories of magnetic shielding: passive and active. Passive shielding was generally applied to magnets produced before 1990; it involves installing large iron plates in the walls, ceiling, and floor of the room that houses the scanner. It is also

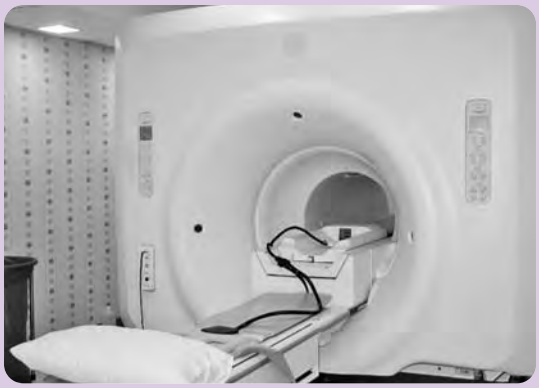


Figure 5. Example of a 1.5 T system with a passive shield design.

Figure 6. Example of a 3.0 T actively shielded system.

possible to directly surround the magnet itself with iron plates, sometimes called self-shielding. The iron plates have the effect of concentrating the magnetic field, so that it is dramatically reduced outside the confines of the shielding structure. The amount of metal required for passive shielding results in enormous weight and greatly increases the load on the floor. While passive shielding is an effective method of containing the fringe field, over the course of approximately 10 years the steel plates themselves can become magnetized and will eventually have to be replaced (Figure 5).

Nearly all new high-field magnets utilize active shielding to restrict the magnet's fringe field. Active shielding in high-field magnets consists of an extra set of windings that surround the coil and generate its own magnetic field. This extra set of windings creates another magnetic field that is the polar opposite of the main magnetic field and which cancels out the main field in the area surrounding the magnet. As a result, the fringe fields are considerably smaller surrounding the system. The reduction in the fringe fields is enough that the 5 Gauss line can easily be located within the scanner room without the need for any additional shielding structure. Active shielding reduces the weight, cost, and complexity of the scanner installation (Figure 6).

RF Shielding

In addition to confining the magnetic field, the design of the scanner room must also minimize interference from RF energy. The signals generated and recorded in MRI are in the same frequency range as those used by radio, television, and other electronic devices. The coils that receive the signal that generates images are sensitive to any source of signals in this frequency range.

RF shielding is used to prevent outside signals from entering the scanner room and interfering with the desired signals for image acquisition. RF shielding can consist of a "cage" of copper sheeting or copper mesh that surrounds the room or the magnet. If built into

the room, it must be applied to the windows and doors of the room as well as to the walls to prevent RF interference from all pathways.

Without proper shielding, external RF energy from common devices found in the hospital setting can generate signals that are as large as, or larger than, the signals generated for imaging. The outside signal source may be a monitoring device, or even a light bulb within the scan room. The result of RF interference are images that are noisy or display characteristic line artifacts in the image, often referred to as a zipper artifact, discussed in another unit in this series.

SUMMARY

High-field scanners remain the most commonly used MRI systems, with 1.5 T field strengths still the most widely used in clinical practice. However, the emerging trend is towards a greater number of 3.0 T systems, particularly in new installations. The growing popularity of 3.0 T systems carries the potential for higher quality and higher resolution imaging, but there are both advantages and limitations to consider when imaging with high-field systems. It is important to realize that 3.0 T systems require extra diligence in following proper screening and facility safety procedures because of their extreme torque (or pull). However, this should not be construed to imply that safety procedures can be more lax at lower field strengths. Low-field systems still function in specialized areas, so an understanding of their applications and capabilities is important for a complete assessment of the most suitable MR examination technique.

COIL TECHNOLOGY

The main magnetic field strength that sets up the alignment of the protons is a major determinant of the quality and SNR the system is capable of providing. The performance of the gradient systems that provide the ability to localize signals and generate images also impacts the speed and resolution possible for the MRI system. The other major subsystem of electronics in an MRI system is the RF system. The RF system is composed of the electronics and coils that generate and record the MR signal that is decoded (reconstructed) to form an image.

BASIC OPERATION OF RF COILS

The RF system performs two primary tasks: exciting the tissue protons by generating a resonating RF signal and converting the altered signals received from the excited tissue into a form used to create the image. Put simply, it sends signal and then receives the signal back in a different form.

B_1 Field Generation

The RF signal transmitted to the body is an oscillating magnetic field at the resonant frequency of the tissue of interest. This oscillating magnetic field is referred to as the B_1 field (recall that the main magnetic field is referred to as the B_0 field.) The first of the two primary tasks of the RF system is to generate an RF signal, a small, oscillating magnetic field that sends energy to the protons in the body tissue such that they begin to spin or precess. This process is called exciting the protons and is accomplished by passing electrical current through the windings of the transmitter RF coil.
The transmitter coil is usually cylindrical, as in the case of the body coil or the transmit/receive head coil.

Receivers and Preamplifiers

The task of the receiver system is to convert the analog signals received by the RF coil into digital form suitable for the computer to use to generate and display an image. A block diagram of the signal path from receiver coil to reconstruction computer is shown in Figure 7.

The signal conversion process is complicated by two factors. First, the signals picked up by the RF coil generate voltages only in the range of microvolts. To be useful, these signals must be greatly amplified. Second, the

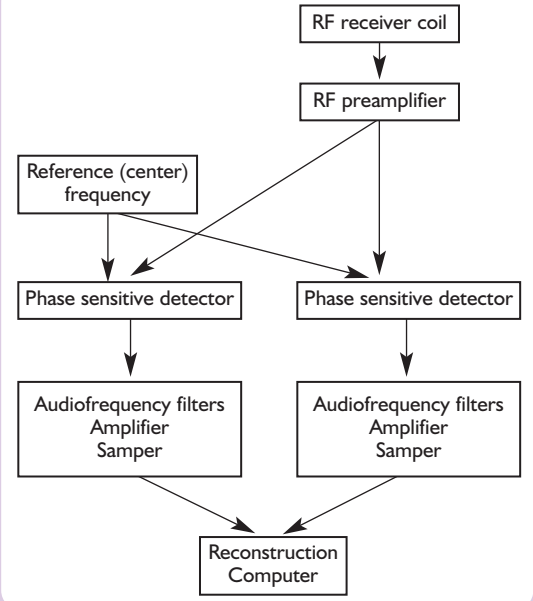


Figure 7. The receiver path signal conversion process is shown beginning at the receiver coil and ending at the reconstruction computer.

signal is oscillating as it is received at the same frequency as the B_1 field generated in the transmit phase, in the radiofrequency range of megahertz. For example, for a 1.5 T system the signals recorded will oscillate at a frequency of about 63 MHz. Filtering to reduce noise and other unwanted signal is difficult in this frequency range. Converting the analog signal to the digital form needed for image reconstruction is also impractical.

The first operation of the receiver system is to amplify the signal generated in the receiver coil to a usable range. This is performed by the preamplifier. The preamplifier strengthens the entire signal received by several orders of magnitude to raise the signal levels from the microvolt range to the millivolt range. One of the most important functions required of the preamplifier is to introduce an absolute minimum of noise into the signal. Otherwise, the noise will be passed along the signal path and greatly degrade the resultant image quality.

The next step is to shift the frequency of the preamplified signal from the RF range of megahertz down to the audio frequency range of kilohertz. The frequency shifting operation is accomplished by mixing the preamplified signal with the original transmitted frequency of the RF excitation. In effect, this process subtracts the transmitted frequency from the received signal. Before the mixing operation, the signals of interest are all in a narrow range around the transmission frequency, so this process converts the oscillation frequency of the signals of interest to a range of several kilohertz (a factor of about 1000) lower than the original. The precise range of frequencies considered is determined by the bandwidth and gradients used during the signal readout. The signals in the resulting lower range are much more convenient to process, because they are in the same range as audio signals and can be processed by computer in much the same way. In particular, the signals can be further amplified, filtered to eliminate noise and other unwanted signals, and digitized to permit computer processing and reconstruction to generate images.

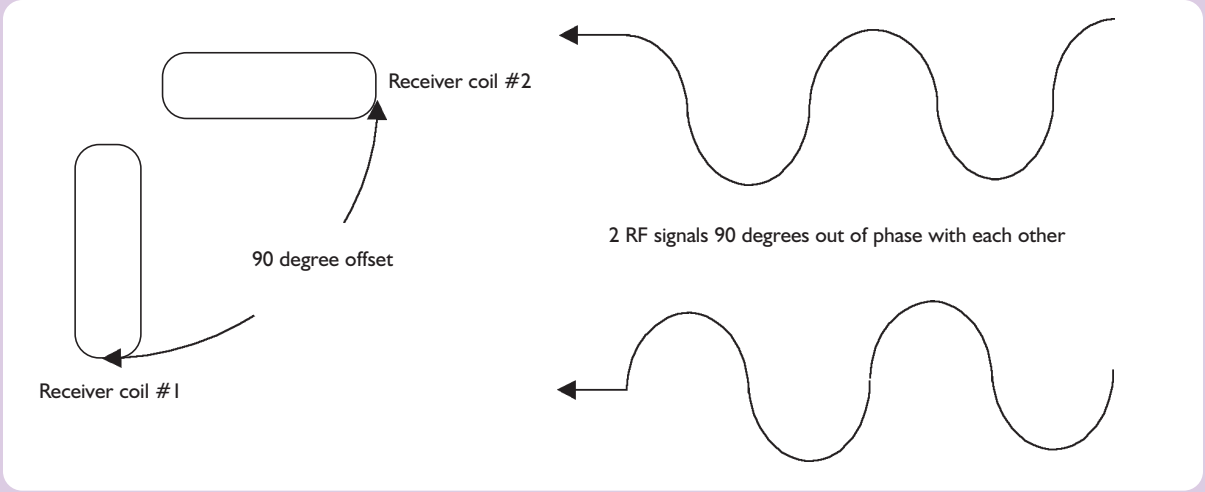


Figure 8a-b. (a) The diagram on the left demonstrates a quadrature coil design that is 90 degrees out of phase. (b) The diagram on the right shows two RF signals. While these signals are the same frequency, the RF pulses are 90 degrees out of phase with each another.

Quadrature detection

In many MRI systems, two detectors are used in the process of converting the signal from its original RF range to the audio frequency range. The two detectors differ by a 90° phase change in the RF signal used for reference with the incoming signal, and the audio frequency range signals are processed separately before image formation. Figure 8 illustrates this concept. The advantage is that the signal output from these two detectors is highly correlated, but the noise in each path is random, resulting in the same effect as two signal averages with respect to the noise. The outcome is an improvement in the SNR by a factor of about 40%.

TRANSMIT AND RECEIVE COILS

Transmit and receive coils use a single coil unit to perform both the excitation (transmit) and the signal readout (receive) functions.

Volume Coils

Common volume coils include the body coil, head coils, and some extremity coils. Volume coils are designed to have a uniform response and SNR over the volume covered, as compared to receive-only surface coils. However, the compromise required for large volume coverage is a reduced SNR over the volume of interest. Transmit and receive volume coils commonly use the quadrature techniques described above to maximize the potential SNR for large volume acquisitions.

Volume coils have several designs:

- **Birdcage coil.** The birdcage coil design is widely used in transmit/receive body coils, head coils, and extremity coils. It consists of a cylindrical arrangement of wires along the longitudinal direction. The arrangement of the wires provides a highly uniform field for RF excitation and signal readout.

- **Saddle coil.** The saddle coil geometry is composed of two loops of a conductor wrapped around opposite sides of a cylinder. The saddle geometry was among the earlier designs for volume coils. It is a simple design, but it suffers from greater RF inhomogeneity when compared to birdcage coils of similar size.

- **Solenoidal coil.** A solenoidal coil is built with closely spaced windings or copper tape on a cylindrical form. Solenoidal coils have a small diameter compared to their length. Solenoidal coils are best suited for use on longer areas of anatomy (greater length) where deep penetration is not required due to the small diameter of the coil. Elbows and knees are examples of anatomy that benefits from the use of solenoidal coils.

RECEIVE-ONLY COILS

Receive-only coils perform only the signal detection function of the MR signal acquisition process. Receive-only coils must be used in combination with a separate transmit coil (usually the body coil) to generate the RF excitation. Because they do not need to transmit RF energy for excitation, receive-only coils can be placed close to the anatomy of interest for imaging and can be much more specialized to achieve the desired anatomic coverage.

Surface coils

Surface coils generally refer to the family of coils that act as receive-only coils and are designed with a specific anatomical part in mind. There are many different types of surface coils, including both single-coil and

multi-coil designs. Surface coils are made in all shapes and sizes, depending on the area of interest.

All other factors remaining the same, the smaller the coil size, the better the SNR. However, the smaller size also means a smaller detection area. Additionally, the depth at which the coil will detect signal is directly proportional to the diameter of the coil; in other words, the smaller the coil diameter, the less the coil can penetrate tissue to capture signal. In particular, the area in which the coil can detect signal will extend to a distance roughly covered by the diameter of the coil element in the lateral directions, and to a depth approximately equal to the radius of the coil. Smaller surface coils are best used for anatomy near the surface of the body because the sensitivity of the coil is greatest near the coil and falls away as the distance from the coil is increased. The closer the coil can be placed to the anatomy to be imaged, the more sensitive the coil will be to that anatomy, and the SNR and image quality will be correspondingly higher.

The advantage of a surface coil is that the coil is sensitive to a considerably smaller local volume, depending on the coil diameter. Both signal and noise are received only in a region of the body near the coil. The higher SNR and reduced coverage means that the anatomy of interest can be imaged with higher spatial resolution. Smaller surface coils are therefore used when high resolution of small structures is required.

The sensitivity of surface coils is not uniform. The signal intensity will be reduced as the distance of the coil from the patient is increased. For surface coils with a single element, this signal variation may cause intensity fluctuations over the anatomy of interest. Proper placement of the coil is important to minimize these effects.

Surface coils come in a variety of shapes and sizes for specific anatomy or for general purpose usage. Examples of specific coils are presented at the end of this material. Some of the general characteristics of large and small surface coils are shown in Table 2.

Surface Coil Correction

The increasing availability of clinical 3.0 T scanners—as well as the development of 8, 16, and 32 receiver systems—has resulted in increased visualization of very high signal intensity in tissue close to the surface coil. In a simple example, a sagittal lumbar spine using a phased array spine coil exhibits extremely bright signal in the posterior fat close to the coil, yet the spinal structures further from the coil and next to the fat are far less bright (phased array coils are described later). The

Large Surface Coils	Small Surface Coils
More uniform sensitivity	Sensitive only near the body surface
Lower SNR	Higher SNR
Not suitable for small FOV	Small FOV and high resolution imaging possible
Applications include chest, torso, hip	Applications include small anatomy such as wrist, elbow

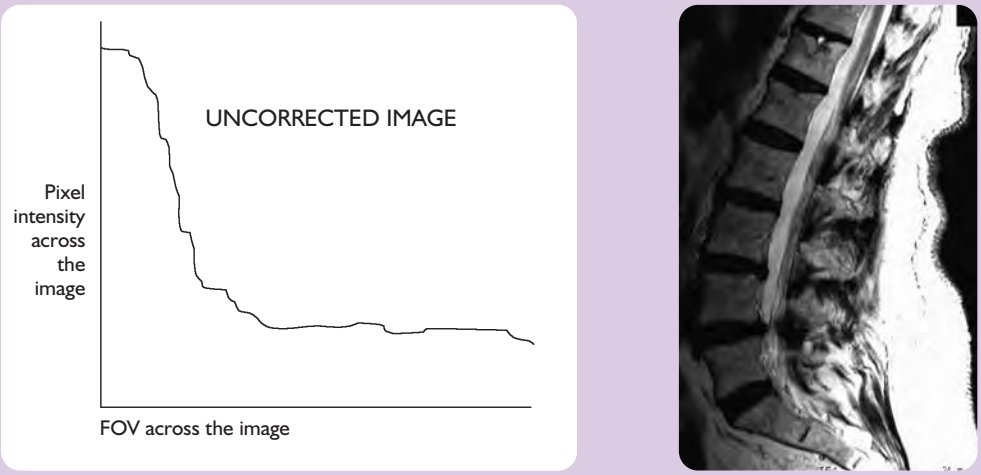
Table 2. General characteristics of surface coils.

great difference in contrast can make it difficult to change the brightness and contrast adjustments in image display for optimal appearance. If the technologist uses window/level for the spinal cord, for example, the posterior fat may be so bright as to be distracting to the radiologist; if window/level is used to "dampen" the extremely bright fat signal, the spinal structures may appear too dark for proper visualization.

Another unsettling example of the phenomenon of nonuniform contrast comes from the increased use of high-element phased array coils (8 or more elements) that completely surround the anatomy of interest. Two examples are a 12-element head array coil and an 18-element chest/torso array coil. The coil design heightens the brightness of all the tissue close to the coil, giving the entire image a haloed appearance, with anatomy closest to the coil brighter than the anatomy toward the center of the image. This appearance can be very distracting to the radiologist, and window/leveling the image to create a more uniform image is virtually impossible. The images that follow offer examples of the signal intensity issues referred to here.

To correct, or at least reduce, the excessive brightness near the coil, MR manufacturers have instituted numerous software-driven programs to correct the intensity. Surface coil intensity correction schemes are implemented either in the reconstruction process or as a postprocessing feature.

The simplest and most common method of surface coil intensity correction postprocessing uses software to view a pixel intensity histogram across the image (see Figures 9 and 10). In the example of the lumbar spine discussed above, the histogram shows very high signal intensity on one side of the image, which then sharply drops off because the signal from the anatomy is farther from the coil. Then the computer reorganizes the image data to reduce the brighter intensities and



a

b

Figure 9a-b. (a) This histogram across the image shown in (b) demonstrates the uncorrected high variation in signal intensity across the image; (b) The sagittal T2-weighted image of the lumbar spine displays characteristic very bright signal from fat posterior and proximate to the spine phased array coil. The extremely bright signal can be visually distracting to the radiologist.

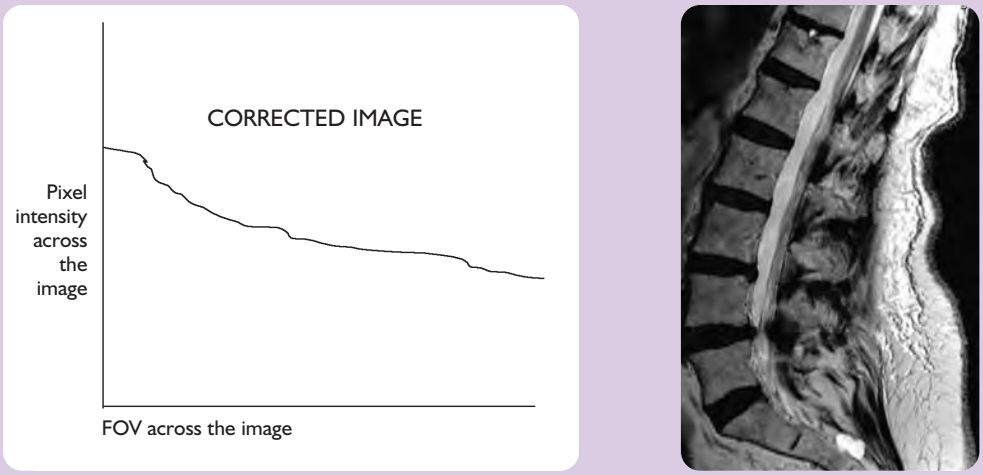


Figure 10a-b. (a) The corrected data corresponding to the filtered image. (b) The same lumbar spine image after correction using an image intensity filter. Note the suppression of the excessive fat signal. The images shown in Figures 9a and 10a have exactly the same brightness and contrast levels.

increase the darker intensities. Finally, the data are reconstructed to display an image with more uniform intensity level across the image.

Phased Array Coils

As mentioned earlier, smaller surface coils yield higher SNR but less coverage. Conversely, a larger surface coil will provide greater coverage, but at the cost of higher SNR.

Initial attempts to create a surface coil with the SNR of a small coil and the coverage area of a large coil used several smaller coils joined together such that they all received signal at the same time. This arrangement is called a multi-coil configuration (Figure 11). Unfortunately, the resulting images yielded only extra coverage, with SNR measurements being equivalent only to a single large coil. The problem with the multi-coil configuration arose from the design, in that all the coils were connected to the same signal receiver.

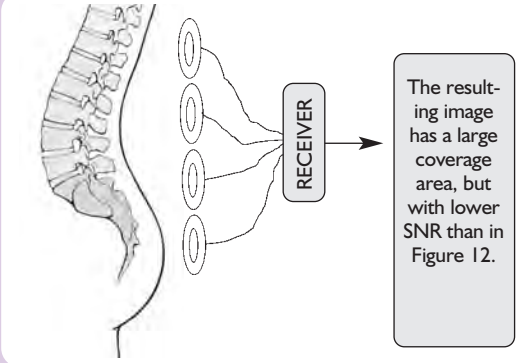


Figure 11 shows a multi-coil configuration, which is not the same as a phased array coil. Multiple coils are connected to a single receiver, so the system reconstructs an image that appears to be from a single large surface coil.

When individual coils were connected to a dedicated receiver the SNR gains were realized, along with the increased coverage. This

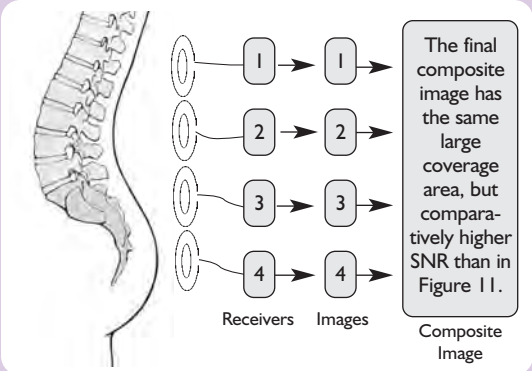


Figure 12 shows a phased array coil configuration. Each coil is connected to its own dedicated receiver, so that each image is processed individually. The individual images are then combined in the last recontruction phase and a single image is displayed, having the SNR of a small coil but the area coverage of a larger coil.

Figure 13. An example of a phased array coil. The photo shows a six-element quadrature cervical-thoracic-lumbar (CTL) phased array coil. The coil elements can be used in any grouping of 2, 3, or 4 coil elements. Typically these combinations are the top 2, 3, or 4 coil elements; the middle 3 or 4 coil elements; or the bottom 2, 3, or 4 coil elements. The particular combination selected is dependent on the area of interest.

discovery in surface coil technology revolutionized MR imaging and is known as a phased array surface coil design (Figures 12 and 13).

Phased array coils are composed of several individual surface coil elements joined into a single unit. Each coil element independently covers a limited volume. The signal from each element is separately recorded and processed; combining the signals from all coil elements creates the final image. This coil design brings the SNR advantages of surface coils to examinations with a larger FOV and anatomic coverage.

The advantages of phased array coils are that the SNR characteristics are the same for each coil element, but the volume that can be covered is greater. The design and arrangement of the coil elements also help to make the final image intensity pattern more uniform. Phased array coils are frequently designed to wrap around the body to cover the chest, torso, or hip with higher SNR than the body coil and more uniform coverage than a single coil element.

SPECIFIC COIL TYPES

Many types of coils have been created for specialized applications to maximize the resolution and image quality for specific types of examinations. Some common types of specialized coils are discussed below.

Head Coils

The brain has always been the most often imaged part of the human body using MRI. Therefore, it is logical that surface coil designs for head/brain imaging are numerous and

diverse. Typically referred to as "head coils," they can be either single channel transmit-receive coils (typically in a birdcage design) or a multi-element, phased array, receive-only coil. The phased array coil designs have various sizes and shapes. Some phased array head coil designs may have the same birdcage appearance as a single-channel transmit-receive coil, but every MR system designates the type of each coil. The technologist is responsible for knowing and selecting the correct coil for any examination based on the specific anatomy being imaged and the SNR/resolution requirements for that type of examination.

Neurovascular Array Coils

One specialized type of head coil is the neurovascular array coil. The neurovascular array coil differs from the head-only coils mentioned above in two areas: the coil is available only in a phased array design, and it has greater length to cover not only the head, but also the neck, cervical spine, and into the mediastinum. The primary use of the neurovascular array coil is to allow imaging of the brain, the intracranial vascular system, and the carotid arteries, all commonly requested examinations. Before the advent of the neurovascular array coil, these examinations required the technologist to change coils during the scan process, switching from the head coil to a cervical or neck coil. This process was time consuming and inefficient. Using the neurovascular array coil, all of the required structures and arteries can be visualized without needing to change coils in the middle of the scan. Moreover, because of the phased array design, the individual coil elements can be selected so that only the brain is imaged for better SNR, or the entire head and neck area can be imaged for complete coverage. An example of a neurovascular array coil is shown in Figure 14.

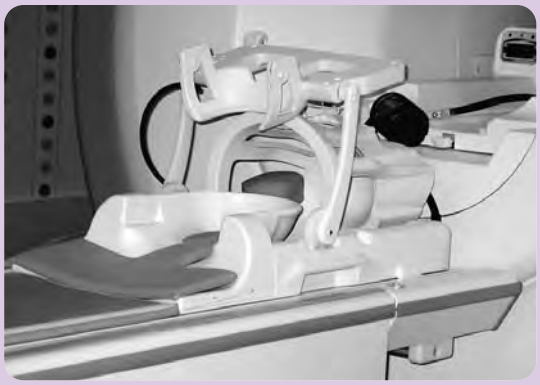


Figure 14. An example of a neurovascular array coil. The coil elements are arranged so the head can be imaged alone or in combination with the cervical spine, soft tissue of the neck, or carotid MRA imaging sequence. The head-only part of the coil can be activated while the neck portion remains inactive if imaging only the brain. When performing carotid MRA, the coverage requirements are from the aortic arch to the Circle of Willis; in this case both the head and neck portions of this coil can be activated together and receive signal in unison.

Spine Array Coils

Second to the head coil, the spine array coil is by far the most often used. The spine array coil is typically long enough to cover the entire length of an average adult spine and contains a sufficient number of coil elements to image any portion of the spine. Depending on the particular manufacturer's design, the entire spine can be imaged in one large FOV, or any portion of the spine may be imaged in a smaller and more detailed FOV.

Muscloskeletal Coils

Many different types of musculoskeletal coils, both phased array and nonphased array, exist for virtually all joints in the body. The most commonly used coils are for imaging the knee, shoulder, and wrist (Figures 15 and 16).

Figure 15. This 3-element array coil is designed specifically for imaging the wrist. The small coil element design ensures high SNR, enabling high resolution images.

Figure 16. This quadrature design transmit-receive coil is used for knee and foot/ankle imaging. Note the column at the top for distal foot imaging and right-angle ankle positioning.

Breast Coils

Specialized breast coils are receive-only coils designed for both unilateral and bilateral imaging of the breasts, as well as of the chest wall and axillary tissue. The most common designs use multiple channels that can be switched on independently for imaging of one or both breasts. State-of-the-art coils are designed with an open architecture to allow access to breasts for patient positioning, needle localization, or MR-guided biopsy.

Peripheral Vascular Array Coil

Complete imaging of the peripheral vasculature for magnetic resonance angiography (MRA)—such as runoff examinations—requires a very large FOV to image from the renal arteries all the way to the feet. Peripheral vascular coils are multiple-element, receive-only coils that permit high SNR imaging of vascular structures. These coils can also be used for soft tissue and musculoskeletal imaging in the lower extremities (Figure 17).

Figure 17 shows a 12-element peripheral vascular array coil designed for use during peripheral run-off MRA examinations. The coil is divided into anterior and posterior halves. The patient lays on the larger posterior half of the coil and the anterior half is placed on top the patient. The coil elements are activated in three groups during the imaging sequence: top, middle, and bottom stations. This coil is designed to provide coverage from the renal arteries to the feet.

The most common configurations of peripheral vascular array coils use two or more components to "wrap" the entire lower body of the patient. The complete patient setup can be performed at the beginning of the examination with no additional adjustments of the coil placement. The multi-element coils improve SNR and uniformity of coverage. Peripheral vascular coils are most suitable for high resolution MRA and bolus tracking studies.

Endocavitary Coils

Endocavitary coils are specialized disposable receive-only coils for very specific clinical indications. Endocavitary coils are used primarily for imaging structures in the pelvic region such as the prostate gland, the cervix, and the colon. For each of these applications, the coil is designed to be positioned inside the rectum, which brings it near to the organ or pathology to be imaged. By bringing the receive element much closer to the organ or tumor, high quality, high resolution images can be acquired over a small FOV (down to about 8 cm) to identify important structures for diagnosis.

The basic design of coils for all endocavitary applications is similar, but the design is individually optimized for prostate, cervical, or colon imaging. For example, the endorectal coil is inflatable and has adjustable deflection for precise positioning closest to the anatomy of interest (Figure 18).

Figure 18. Endorectal coils.

SPECIALIZED ARRAY COILS FOR PARALLEL IMAGING

Parallel imaging has been a breakthrough technology for high speed imaging. Parallel imaging has the potential to dramatically improve scan speeds for many types of routine clinical examinations and make higher resolution possible in a reasonable scan time.

Parallel imaging is also becoming important for 3.0 T imaging to overcome the limitations imposed by FDA-regulated SAR. SAR is the maximum amount of RF energy that can be transmitted to the patient within a designated amount of time according to the patient's weight as defined by the FDA. Parallel imaging uses multiple-element coils (phased array coils) for signal reception and multiple receiver channels to independently process the image data from each coil element. The data from the individual coil elements are then combined to reconstruct the final image.

Concept and Theory of Operation

The fundamental concept of parallel imaging is based on the fact that the different elements of a multiple-element receiver coil will have different sensitivities to a particular location in the body. The amplitude of the signal produced in a coil element from any specific location in the body decreases as the distance from that location increases. Historically, this change in amplitude was considered a draw-

back of multiple-element coils because it introduces a nonuniform intensity variation over the FOV; however, this phenomenon can be exploited to allow much faster imaging.

The key to parallel imaging is to recognize that spatial differences in coil sensitivities can be used as a way of encoding spatial information. Each voxel of tissue will generate a different signal in each coil element, depending on the position of the tissue relative to that coil. By comparing the responses in each of the coils, some information about the spatial location of the tissue can be generated. The tissue will generate more signal in the coil elements it is closer to, as compared to the coil elements farther away. Therefore, not only is spatial information generated by frequency and phase-encoding gradients but also from the arrangement of and signal in the coil elements. Special reconstruction software can extract the spatial information from the signals arising from the independent coils and use it to replace some of the information that would normally be acquired by phase encoding. The end result is that all of the needed spatial information can be acquired using fewer phase-encode lines, which means that the acquisition time is shortened. Ideally, the number of phase-encode lines could be reduced by a factor equal to the number of coil elements, but in practice the reduction factor that can be achieved is somewhat lower. The trade-off for a shortened acquisition time is a lower SNR.

To apply parallel imaging, the particular sensitivity pattern in each of the coils must be known. The sensitivity information can be taken either from the known and fixed design of the coil or (more commonly) from a low resolution calibration scan that estimates the sensitivity pattern during the examination. These data are needed during the image reconstruction to reduce the acquisition time and still generate a full FOV image.

Although parallel imaging can accelerate an imaging examination by a factor equal to the number of coil elements in the array coil (since each coil element provides unique spatial information), in practice this acceleration factor is rarely achieved. Additionally, there are trade-offs that must be considered when determining what acceleration factors can be used to produce a high quality examination.

The biggest drawback in a parallel imaging acquisition is the reduction in SNR as compared to a fully sampled scan. SNR is related to the total duration of the signal encoded by the readout gradient. Thus, if an acceleration factor of four is used for a parallel acquisition, a loss of a factor of at least two can be expected in SNR; this decrease is in addition to the SNR loss related to the coil geometry. The loss of SNR coupled with the dramatic increase in imaging speed makes parallel imaging perfect for applications like body imaging, where FOV and slice thickness yield sufficient SNR to allow parallel imaging without a prohibitive loss of SNR. Musculoskeletal imaging applications may not be good candidates for parallel imaging, since they are typically very high resolution (small FOV and thin slice thickness) and therefore already tend to experience low SNR. Nevertheless, many types of acquisitions (such as FSE techniques) have sufficient SNR that even with the SNR loss due to the reduced scan time and coil geometry, high quality images are still acquired in a much shorter time, permitting higher resolution, or reduced motion and other artifacts.

Special Coil Design Issues

Since it is the arrangement of coil elements—and their relative responses—that make parallel imaging possible, it is not surprising that the geometry of the coils is very important in achieving maximum benefit from parallel imaging. Theoretically, the number of coil elements determines the maximum amount of acceleration possible. However, in an actual coil design there will be some overlap of the areas covered by the individual coil elements, and the information acquired from each coil will not be completely independent. This means that using the maximum acceleration factor (of the number of coil elements) will likely leave some artifact in the resultant images.

Also, because the coil response is not uniform, noise in the images will be amplified in some areas in the final image reconstruction. The amount of increased noise and its distribution in the FOV is measured by the geometry factor (or g-factor) of the coil arrangement. Ideally, the g-factor would be equal to one, indicating no additional noise enhancement from the coil arrangement. The g-factor depends very strongly on the exact geometry of the coils and usually must be measured with simulations. Careful attention to the coil arrangement can minimize the g-factor and permit acceleration factors close to the maximum (Figure 19).

Vendor-specific Implementations

Not all MRI manufacturers offer a parallel imaging pulse sequence, but those that do use a vendor-specific name. For example, Philips Healthcare offers two types of parallel imaging known as SENSE and SMASH, although SENSE is by far used more often. GE Healthcare offers two sequences known as ASSET and ARC, and the Siemens sequence is known as iPAT. Although each sequence has slight differences in design and implementation, all parallel imaging techniques abide by the same basic principles and with the same advantages and pitfalls.

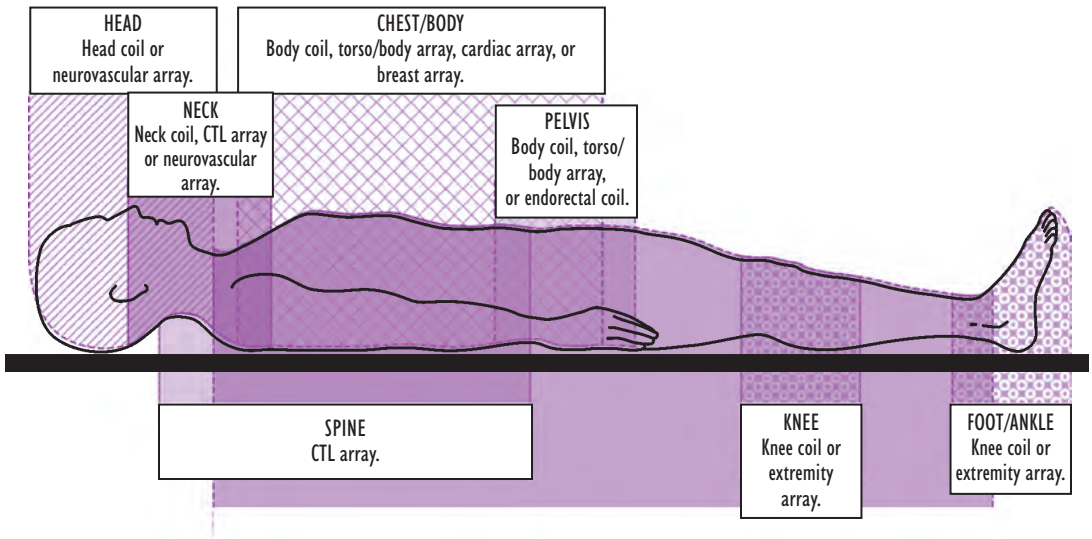


Figure 19 demonstrates appropriate body coverage using the different types of coils.

SUMMARY

Surface coils are an extremely integral part of any complete MR system. The coils, as well as the entire RF subsystem, are as important as any other major component and as important as the gradients and even the magnet itself. Advances in coil technology and in the entire RF chain have contributed significantly to MRI's exponential growth in applications and reductions in scan time while improving image quality. Continued technological advancements in the area of number of receiver channels seed the development of new high-channel phased array coils. The continued evolution of clinical ultra high-field imagining, such as 3.0 T and higher, will also foster new developments in surface coil technology.

Figures

1	Courtesy of Hitachi, Ltd.
2a	Courtesy of GE Healthcare
2b, 6	Courtesy of Siemens Medical Systems
2c	Courtesy of Philips Medical Systems
4	Tom Schrack, BS, ARMRIT
5	Courtesy of Fairfax Radiology Consultants
6	Courtesy of Siemens Medical Systems
7	Dan Thedens, PhD and Alan Stolpen, MD, PhD; adapted from Hinshaw WS, Lent AH. *An Introduction to NMR imaging: From the Bloch equation to the imaging equation.* Proceedings of the IEEE 1983;71:338-350.
8, 9b, 10b, 11-12	Tom Schrack, BS, ARMRIT
9a, 10a, 13-17	Courtesy of Fairfax Radiology Consultants
18	Courtesy of Medrad, Inc.
19	Adapted from Siemens Medical Solutions and *MRI for Technologists, Module 2: Technical Considerations of MRI*, ©2002

Tables

1	Dan Thedens, PhD
2	Dan Thedens, PhD and Alan Stolpen, MD, PhD

Additional Reading

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ABBREVIATIONS OF TERMS

ADC	apparent diffusion coefficients
AUC	area under the curve
B_1	the RF transmission field
B_0	the main magnetic field
bFFE	balanced fast field echo
BOLD	blood oxygen level dependent
cm	centimeter
CNS	central nervous system
CSF	cerebrospinal fluid
CT	computed tomography
CTL	cervical-thoracic-lumbar
dB/dt	time-varying B fields (gradient-altered B_0)
dB	change in the main magnetic field (B_0) –OR– decibel
dBA	decibel attenuation
DSC	dynamic susceptibility contrast
DTI	diffusion tensor imaging
eADC	exponential ADC map
EKG	electrocardiogram
EPI	echo-planar imaging
FLAIR	fluid attenuated inversion recovery
***f*MRI**	functional magnetic resonance imaging
FNH	focal nodular hyperplasia
FS	fat suppressed
FSE	fast spin-echo
FOV	field of view
G	Gauss
g-factor	geometry factor
GRE	gradient echo
Hz	hertz
IR	inversion recovery
IV	intravenous
kHz	kiloHertz (1,000 hertz)
KVO	keep vein open
mHz	megaHertz (1,000,000 hertz)
min	minute(s)
MinIP	minimum intensity projection
MIP	maximum intensity projection
mm	millimeter
mmol/kg	millimole per kilogram
mOsm/kg	milliosmoles per kilogram
mOsm/L	milliosmoles per Liter
MPR	multiplanar reconstruction
MRI	magnetic resonance imaging
MRA	magnetic resonance angiography
MRCP	magnetic resonance cholangiopancreatagraphy
MRS	magnetic resonance spectroscopy
MRV	magnetic resonance venography
mT	milliTesla
mT/m	milliTesla per meter
mT/m/s	milliTesla per meter per second
MTT	mean transit time
NAA	N-acetyl aspartate
NEX	number of excitations (also NSA)
nm	nanometer
NSA	number of signal averages (also NEX)
OCP	o-cresolpthalein complexone technique
Osm/kg	osmoles per kilogram
PACS	picture archiving communication system
PNS	peripheral nerve stimulation
r_1	T_1 recovery
r_2	T_2 decay
RBW	receive bandwidth
rCBV	relative cerebral blood volume

RES	reticuloendothelial system
RF	radiofrequency
ROI	region of interest
SAR	specific absorption rate
SE	spin-echo
SNR	signal-to-noise ratio
SPGR	spoiled gradient echo
SPIO	superparamagnetic iron oxide
SSFP	steady state free precession
STIR	short tau inversion recovery
T	Tesla
T1	time for 63% of a tissue's longitudinal magnetization to recovery
T2	time for 63% of a tissue's transverse magnetization to decay
TE	echo time
T/m/s	tesla per meter per second
TOF	time of flight
TR	time to recovery –OR– repetition time
TSE	turbo spin-echo (Siemens Medical Systems term for fast spin-echo)
UAE	uterine artery embolization
USPIO	ultrasmall superparamagnetic iron oxide
W/kg	watts/kilogram

GLOSSARY

absolute zero
the theoretical point at which all molecular motion stops, measured as 0^0 Kelvin

aliasing
a common artifact caused when the FOV selected is smaller than the area of tissue excited; also known as "wrap-around"

algorithm[1]
a step-by-step method of solving a problem or making decisions, as in making a diagnosis. An established mechanical procedure for solving certain mathematical problems

artifact[1]
in radiology, a substance or structure not naturally present in living tissue, but of which an authentic image appears in a radiograph

BOLD (blood oxygen level dependent) imaging
an imaging technique based on the EPI pulse sequence in which the patient is given mental tasks to perform while undergoing MR examination. Increases in blood flow to affected areas of the brain are postprocessed into functional maps indicating brain activity.

bore
the tubular portion of the magnet in which the patient is placed

contraindication
an absolute reason or cause not to proceed with a diagnostic examination or procedure. In MRI it is typically a patient condition, such as an embedded cardiac non-MR compatible pacemaker, that prohibits the patient from undergoing MR examination.

contrast
differences in signal intensity between two adjacent areas on an MR image

cryogen
super-cooled liquid gas used to cool a given material to near absolute zero, thus becoming superconductive; in MR, liquid helium

cryostat
the vessel in which the magnet is immersed in liquid helium

diffusion imaging
an imaging technique usually based on the EPI pulse sequence; it is used to indicate the amount of water absorption across a brain cell membrane

duty cycle
the amount of time (stated in percentage) that an MR system gradients can be "active" before they must be turned off in order to cool; most MR manufacturers report a duty cycle of 100%, indicating the gradients can run at full power continuously

echo planar imaging (EPI)
a very fast pulse sequence characterized by rapid oscillation of the frequency encoding gradient to create an echo train; EPI fills k-space quickly

echo spacing
the time from the peak of one echo in an echo train to the next

echo train
the series of echoes created by a FSE or EPI pulse sequence

electromagnetic spectrum
continuous series of different types of electromagnetic radiation, ordered according to wave-length of frequency

fast spin-echo (FSE)
a rapid pulse sequence characterized by a series of 180^0 RF echo pulses used to create numerous echoes within a single TR, thus filling k-space more quickly

fat suppression
any of the methods used to reduce signal from fat on an MR image

ferromagnetic
materials that react to a magnetic force; all iron and some stainless steel are ferromagnetic

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

fluid attenuated inversion recovery (FLAIR)
inversion recovery-based pulse sequence that utilizes a relative long TI in order to suppress signal from a long T2 tissue while maintaining heavy T2-weighting throughout the rest of the image

flip angle
the rotation of the amount of RF energy used to excite some portion of protons in the longitudinal plane into the transverse plane

frequency
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

fringe field
the extended magnetic field generated by an MRI magnet system that is outside the magnet bore, or scan area. The fringe field decreases in strength as the distance from the magnet increases

functional MRI (*f*MRI)
any of the imaging techniques that demonstrate function of anatomical structure; examples include spectroscopy, dynamic liver imaging, cardiac imaging, and BOLD imaging

Gauss[1]
[Karl Friedrich *Gauss,* Geman mathematician and physicist, 1777-1855] the cgs unit of magnetic flux density, equal to 10^{-4} tesla. Symbol, G

gradient (magnetic field)
a magnetic field that changes in strength along a given direction

gradient amplitude
the degree to which a gradient field can vary from zero to the peak maximum point; typically measured in milliTesla/meter (mT/m)

gradient coil
one of six coils (3 pair) placed in the three orthogonal planes (denoted X,Y, and Z) that generate small magnetic fields along the plane of the main magnetic field; it is used for slice selection, phase and frequency encoding

gradient echo
pulse sequence characterized usually by a $<90^0$ RF excitation pulse and an echo generated by gradient reversal instead of a 180^0 RF echo pulse

gradient moment nulling
a method to reduce flow artifact in which one, or all three, pair of gradients are pulsed to dephase and rephase spins that flow along the axis of that gradient

homogeneity
the uniformity of any field; in MRI it is the uniformity of the B_0 field

inhomogeneity
absence of homogeneity or uniformity; inhomogeneity in a magnetic field may occur as one area of the field deviates from the average magnetic field strength

inversion recovery
pulse sequence characterized by an initial 180^0 RF inversion pulse

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 can begin

Larmor frequency
the frequency at which magnetic resonance is produced in a sample of hydrogen nuclei, or other types of nuclei used in MRI

ligand[1]
a molecule that donates or accepts a pair of electrons to form a coordinate covalent bond with the central metal atom of a coordination complex

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

magnetic shielding
metal surrounding an MR magnet used to contain the main magnetic field fringe field within acceptable limits; magnetic shielding can be passive (steel lined walls) or active (built into the system)

magnetic susceptibility
the degree to which a tissue can become magnetized

maximum intensity projection (MIP)
a ray tracing algorithm where a ray goes through a designated imaging block or volume. Signal intensity is designated based on nearness to the observer

motion artifact
an artifact or signal not naturally present in living tissue, but which appears on MRI film due to movement of muscle or fluid or motion of any body part

multiplanar reconstruction (MPR)
two-dimensional views of a single voxel thickness of vascular structures reconstructed from three-dimensional or multi-slice images

magnetic resonance angiography (MRA)[1]
a form of magnetic resonance imaging used to study blood vessels and blood flow, particularly for detection of abnormalities in the arteries and veins throughout the body

magnetic resonance imaging (MRI)
a method of visualizing soft tissues of the body by applying an external magnetic field that makes it possible to distinguish between hydrogen atoms in different environments

magnetic resonance spectroscopy (MRS)
MR examination in which the data collected are not reconstructed into an image, but into a spectrum of signals based on metabolite presence within the tissue being examined; a type of *f*MRI

nanometer (nm)[1]
a unit of linear measure equal to one-billionth of a meter, 10^{9} meter

number of excitations (NEX)
the number of cycles of completed k-space filling; also known as number of signal averages (NSA)

number of signal averages (NSA)
see number of excitations

paramagnetic
a substance with magnetic properties that may significantly reduce T1 and T2 relaxation times in MRI

perfusion imaging
an imaging technique based on the EPI pulse sequence; gadolinium contrast is used to indicate the amount of blood perfusion across a brain cell membrane

peripheral nerve stimulation (PNS)
activation of a peripheral nerve fiber(s) caused by rapidly switching gradient fields; PNS is not a patient safety concern but a potential patient comfort concern

phase
particular stage or point of advancement in a cycle

phased array coil
a type of surface coil composed of several coils and receivers that are linked together. The signals from each of the coils and receivers are subsequently united to form an image with good SNR.

phase encoding
generation of phase differences along a particular direction of a tissue slice for use in spatial localization of MR signal

pixel
picture element; the smallest discrete part of a digital image display

pulse sequence
a series of events for exciting protons and detecting signals during the MR examination; every pulse sequence includes slice excitation, echo generation, and phase and frequency encoding

quench
sudden and massive expansion of liquid helium into gaseous helium due to an increase of heat from a loss of superconductivity of the magnet

R-R interval
the period of time between each R-wave; one cardiac cycle

R-wave[1]
the initial upward deflection of the QRS complex, following the Q-wave in the normal electrocardiogram

radiofrequency (RF)[1]
the range of frequencies of electromagnetic radiation between 10 kilohertz and 100 gigahertz that is used for radio communication

ramp time
the minimum time required for a gradient field to go from its peak maximum point to its peak minimum point; measured in microseconds

ramp up/ ramp down
the controlled process of bringing the magnet to maximum field or reducing the magnet from its maximum field

receive bandwidth (RBW)
the range of frequencies collected during the frequency encoding portion of the pulse sequence

region of interest (ROI)
a specific defined area, ie, fluid or a portion of an organ or tumor, where a relative signal intensity measurement can be obtained

RF shielding
metal used to prevent stray RF frequencies from entering the magnet room during an MR exam, typically made of copper

rise time
the minimum time required for a gradient field to go from zero to its required maximum; measured in microseconds

resonance
state of a system through which energy may be transferred to another system with the same preferred or resonant frequency; characterized by absorption and dissipation of energy through resonant oscillation

saturation
lack of signal due to overexcitation of protons

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

slew rate
describes overall gradient performance as a function of gradient amplitude and gradient rise time. Slew rate is derived by dividing the amplitude by the rise time and typically is described in units of T/m/sec.

specific absorption rate (SAR)
the FDA-regulated amount of RF heat energy that a patient can absorb during an MRI exam, measured in Watts/kilogram

spin-echo
basic pulse sequence of MR imaging using a 90^0 RF excitation pulse and a 180^0 RF echo pulse

static magnetic field
the large main magnet field generated by the magnet to place the protons into the longitudinal plane prior to the MRI pulse sequence; also known as the B_0 field

short tau inversion recovery (STIR)
inversion recovery-based pulse sequence that utilizes a relatively short TI time in order to obtain heavy fat suppression; the TI is selected based on T1 recovery time of fat

superconductivity
the state where all molecular motion becomes so restricted, due to a lack of heat, that electricity can flow through a conductor without resistance. Materials such as a magnet coil become superconductive after being immersed in liquid helium and reaching a temperature of 4^0 Kelvin

surface coil
specialized antenna for transmitting and/or receiving RF energy to or from the patient during a pulse sequence

T1
the amount of time for 63% of a tissue's protons to recover to longitudinal magnetization

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

T2
the amount of time for 63% of a tissue's protons to decay in the transverse plane

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

Tesla[1]
the SI unit of magnetic flux density, calculated as webers per square meter. It replaces the gauss.

time to echo (TE)
the amount of time selected by the technologist to allow for T2 decay of excited protons; also the amount of time from the beginning of initial slice excitation pulse and the generated echo signal

time-of-flight imaging (TOF)
gradient echo-based pulse sequence that uses flow-related enhancement to greatly increase the contrast between flowing blood and stationary tissue; performed in either a 2D or 3D acquisition, the images are postprocessed into a maximum intensity pixel projection (MIP) to display like an MR angiogram

time to inversion (TI)
the time allowed from the initial 180 degree excitation pulse until the 90 degree pulse in an inversion recovery pulse sequence; TI determines the amount of T1 recovery time for a given tissue

torque[1]
a rotatory force causing a part of a structure to twist about an axis

time to recovery (TR)
the amount of time selected by the technologist to allow for T1 recovery of the excited protons; also the time from the beginning of one pulse sequence to the beginning of the next

trigger window
the time delay before each R-wave

vector
mathematical quantity representing both magnitude and direction, symbolized by an arrow

voxel
a three-dimensional volume of tissue corresponding to a pixel on an MR image, a "volume element"