Nuclear magnetic resonance (MR) is a physical phenomenon that has been exploited for many applications in physics and chemistry since the late 1940s. It has become increasingly important in recent years as the basis of a relatively new medical imaging technique, MR imaging. In particular, MR imaging promises to become a valuable tool for the evaluation of cardiovascular disease. This article briefly reviews some of the basic aspects of the phenomenon of nuclear MR and how they are applied in MR imaging, with particular reference to cardiovascular applications. Because only a broad overview of the basic principles is provided, specific references to the literature are not made.

NUCLEAR MAGNETIC RESONANCE

Nuclear Magnetization

The nuclei of certain atoms, including hydrogen (which is ubiquitous in the body in many molecules, including water and fat), exhibit magnetic properties. That is, they tend to line up with an externally applied magnetic field. This results in a net bulk magnetization (albeit a weak one) of the tissue and fluids of the body along the direction of the magnetic field (longitudinal magnetization) when placed in a strong magnetic field. The magnetic fields used in MR imaging systems are on the order of 10,000 times or more stronger than the earth’s magnetic field. Although these strong magnetic fields are not in themselves harmful, they may exert forces on any magnetic foreign bodies, such as some intracranial aneurysm clips, or they may affect some implanted medical devices, such as cardiac pacemakers.

When the body is placed in a magnetic field, the magnetization of the nuclei is not instantaneous but builds up to an equilibrium value with a characteristic time constant, called the TI relaxation time, which depends on the particular tissue and its state (Fig. 1). For example, myocardium has a longer T1 time than fat, and the T1 of blood is longer than that of myocardium; the T1 of edematous tissue is longer than that of normal tissue. The magnetization can also be altered during imaging; again, it recovers toward its equilibrium value at a rate determined by the T1 relaxation time. This dependence of the magnetization (and thus the nuclear MR signal) on the local relaxation time is an important source of image contrast in MR imaging.

Nuclear Resonance

Magnetized nuclei also exhibit the phenomenon of magnetic resonance. That is, there is a particular frequency (proportional to the
external magnetic field) at which they interact strongly with an (additional) oscillating magnetic field. This frequency is characteristic of the particular type of nucleus. For example, for the nuclei of hydrogen (protons), the resonance frequency is 42.58 MHz/T. (T stands for tesla, equal to 10,000 gauss. The gauss is an older unit of magnetic field strength that may be more familiar; the earth’s magnetic field is typically on the order of half a gauss.) Because most MR imaging systems have magnetic field strengths on the order of a tesla, their operating frequencies are in the radiofrequency (RF) range.

In their equilibrium state, the magnetic nuclei act collectively to produce a net bulk magnetization oriented along the external magnetic field. The bulk nuclear magnetization is weak and is difficult to detect in this static equilibrium condition. When an external oscillating magnetic field is applied at the resonance frequency, the orientation of the nuclear magnetization can be rotated away from its equilibrium state (Fig. 2). The net amount of rotation of the nuclear magnetization vector depends on the strength and duration of the applied resonance frequency magnetic field. Thus the magnetized nuclei (spins) can be rotated any desired amount (e.g., 90 or 180 degrees) by applying a suitable pulse of resonance frequency magnetic field. The amount of the rotation of the magnetization vector is termed the flip angle.

Perturbing the spins away from their equilibrium state is called excitation. As mentioned earlier, the resulting loss of the component of the bulk magnetization along the external magnetizing field is recovered back toward the equilibrium condition at a rate determined by the local T1 relaxation time. In general, however, the process of excitation also results in creation of a vector component of the nuclear magnetization perpendicular to the external field (transverse magnetization). This transverse component now rotates around the external field (precesses) at the resonance frequency. This rotating magnetization produces a weak but readily detectable voltage in a suitable receiver coil external to the body; this is the nuclear MR signal. The transverse magnetization decays away at a rate determined by a second characteristic relaxation time constant, the T2 relaxation time (Fig. 3). The T2 time also depends on the...
tissue type and state. It is shorter than or on the order of (for liquids) $T_1$ and is another important determinant of image contrast.

**Signal-to-Noise Ratio**

The MR signal is weak and can easily be swamped by superimposed random voltages (noise). To maximize the signal-to-noise ratio (SNR), the MR imaging system is enclosed in a RF-shielded room, and all the associated electronics are designed to be low noise. A large portion of the noise superimposed on the signal actually comes from the body itself. To minimize this noise, receiver coils designed to be sensitive to a relatively restricted portion of the body can be used, although this correspondingly restricts the field of view of the imaging. For a given level of noise, the smaller the region from which the signal is gathered (e.g., the higher the image resolution), the weaker the signal coming from that region will be, and the poorer the resulting SNR will be. This can be a limiting factor in practically achievable image resolution. Acquiring imaging data repeatedly and averaging it can improve the SNR by a factor of the square root of the number of averages but correspondingly prolongs the time required for data acquisition.

**Chemical Shift**

The resonance frequency of a nucleus is determined by the local strength of the magnetic field at the nucleus. At different positions within a molecule, the nuclei experience slightly modified magnetic field strengths, on the order of a part per million (ppm) difference. For example, the hydrogen in fat resonates at a frequency that differs by about 3 ppm from the hydrogen in water. These differences in resonance frequency have been used by chemists for decades to study molecular structure and chemical composition. They can also be used to study in vivo biochemistry noninvasively. For example, MR spectroscopy of phosphorus allows detection of metabolically important molecules such as adenosine triphosphate (ATP) and phosphocreatine (PCr), although not with as good an SNR as conventional hydrogen imaging.

**Spin Echo**

In the equilibrium state, the nuclear magnetizations are all aligned together, along the external magnetic field. When transverse magnetization is created, the nuclei are initially all still aligned so that their bulk magnetization processes together, producing a signal with a particular phase (e.g., the difference between a sine wave and a cosine wave). Because of local variations in the magnetic field (and possible chemical shift differences), however, the individual nuclear magnetizations tend to get out of phase with each other, so that the observed signal decays with a shorter time constant than the intrinsic $T_2$ relaxation time, the $T_2$ relaxation time. The full signal strength, limited only by the $T_2$ time, can be transiently recovered as a spin echo by applying a second pulse of resonance frequency magnetic field, typically set to produce 180 degrees rotation of the nuclear mag-
netization, at some delay time after the initial excitation. This second excitation pulse can cause a transient resynchronization of the phases of the nuclear magnetizations and a corresponding transient buildup of the signal to the strength determined by the T2 relaxation time.

**Contrast Agents**

The relaxation times can be shortened in the presence of certain chemicals. In particular, certain chelates of gadolinium, a rare earth element with strong magnetic properties, are inert and well tolerated and can be given intravenously. These gadolinium contrast agents shorten the relaxation times of tissues where they accumulate (reflecting such factors as vascularity and extravascular exchange), thus altering the local intensity in MR images. The dynamic observation of the passage of a contrast agent bolus through a tissue may help assess regional perfusion. Alternatively, imaging early after the injection, while the contrast agent is still concentrated in the blood, can bring out images of the blood vessels and cardiac chambers.

**MAGNETIC RESONANCE IMAGING**

**Magnetic Field Gradients**

The magnets in MR imaging systems are designed to produce a uniform field over the region to be imaged. Thus, ideally, all nuclei have the same resonance frequency. Because the external signal receiver coil has only limited localizing ability, there would be no direct way to make useful images from the detected signal. In addition, if all the nuclei have the same resonance frequency, there is no way selectively to excite only a desired region (slice) to be imaged.

These limitations can be overcome with the provision of the capability to create local magnetic field gradients (gradients) within the region to be imaged (Fig. 4). These are generated with suitable arrangements of current-carrying conductors, designed to generate magnetic fields that vary linearly over the region to be imaged. A set of three such gradient-generating systems, designed to produce magnetic field gradients in orthogonal directions, can be used together to create net gradients in any desired direction.

**Selective Excitation**

When used during excitation with an oscillating magnetic field at a given frequency, the gradient magnetic field locally adds or subtracts from the steady external magnetic field, so that only a desired plane perpendicular to the direction of the gradient is exactly on resonance and only a thin slab centered on this plane is excited (Fig. 5). Thus, gradi-
ents can be used to create selective excitation of only a desired slice for imaging. Because the gradient orientation direction can be freely chosen, this permits clinicians correspondingly to image in any desired orientation.

**Frequency Encoding**

A magnetic field gradient can be created along a desired direction in the excited region during MR signal detection. This allows one to relate the frequency of the detected signal to its position along the direction of the gradient, through the linear relationship between local magnetic field strength and corresponding resonance frequency. In general, there are a range of signal frequencies, corresponding to an extended object to be imaged along the direction of the gradient, whose shifting relative phases over time combine to cause a time-varying signal that can be detected. The mathematical operation called the *Fourier transform* can be used to calculate the equivalent distribution of signal strengths at different frequencies, from the observed variation of the signal at different times (e.g., in the presence of a steady magnetic field gradient) (Fig. 6). These values can then be used to calculate the corresponding projection of signal intensities in the excited slice onto the direction of the gradient, through the known applied gradient strength and orientation. Applying a preliminary pulse of magnetic field gradient in the opposite direction from the frequency encoding gradient leads to initial suppression of the signal because of the resulting signal phase variation along the gradient direction. Reversal of this phase variation along the gradient direction during the signal detection in the presence of the frequency encoding gradient results in a transient reappearance of the signal as a gradient echo.

**Phase Encoding**

As described previously, a frequency encoding gradient during signal detection can be used to find the projection of the signal distribution onto the direction of the gradient. This projection can be used to reconstruct an image by acquiring a series of such signal data with successive rotations of the direction of the gradient, analogous to x-ray computed tomography. Most MR imaging systems instead use phase encoding gradients, however, that is, a series of pulses of magnetic field gradient of variable strength, oriented orthogonally to the frequency encoding gradient, which are applied before a corresponding series of signal detections in the presence of a frequency encoding gradient. With a consistent strength and direction of the frequency encoding gradient, the different phase distributions that evolve along the direction of the phase encoding gradient with different strength pulses result in different patterns of signal interference and corresponding modulation of the calculated signal projections. With a set of such signals acquired with a suitable set of phase encoding gradients, the Fourier transform can be used to calculate the signal strength distribution in the direction of the phase encoding gradient and thus reconstruct a full image.

**Image Contrast**

An important determinant of image intensity is the density of signal-producing nuclei
(e.g., aerated lung is much darker than most other tissues). Most tissues, however, have similar density. Thus, clinicians instead rely on relaxation time differences for most useful image contrast. The delay between initial excitation and signal detection (the echo time, TE) allows some decay of the transverse magnetization to take place. For a given value of TE, the shorter the local value of T2, the greater the amount of this signal decay and the darker the region will appear in the final image. Imaging with a longer value of TE results in a greater relative amount of this T2-weighting effect, in which regions with longer T2 tend to appear relatively brighter.

Because multiple acquisitions with different values of the phase encoding gradient are generally necessary to generate an image, the excitation process must be repeated multiple times. If the interval between successive excitations (the repetition time, TR) is on the order of or shorter than the local T1 relaxation time, the recovery of the nuclear magnetization toward its equilibrium value will be incomplete (it will be partially saturated), and a weaker signal will result from the subsequent excitations. Imaging with a shorter value of TR results in a greater amount of this T1-weighting effect, in which regions with longer T1 tend to appear relatively darker.

Preconditioning excitation pulses can be used before the actual signal excitation pulses to manipulate image contrast. For example, if signal acquisition is preceded with a combination of a 180-degree (inversion) pulse to invert the magnetization and an intervening recovery interval, the final T1 contribution to the image contrast can be influenced through the varying amounts of remagnetization during the inversion recovery time (TI).

FLOW EFFECTS

An important aspect of cardiovascular applications of MR imaging is its sensitivity to flow and motion effects. Flow effects can be used to enhance the contrast between stationary tissue and flowing blood. They can also be used to measure blood velocity. They can even be used to study regional myocardial motion. Conceptually, flow effects can be grouped into time-of-flight effects and phase shift effects. Time-of-flight effects result from the fact that if the magnetization of a tissue is locally perturbed and the tissue then moves, the local perturbation of magnetization moves with the underlying tissue and persists for times on the order of the relaxation times (Fig. 7). For example, in conventional T1-weighted imaging, repeated excitation of the region being imaged results in relative suppression of the signal from stationary tissue. If blood flow can carry in more fully magnetized spins from outside the imaging region between excitations, however, the blood vessels may appear relatively brighter than the adjacent stationary tissue. Rapidly repeated excitations (short TR), with small flip angles and signal detection as a gradient echo, can be used to produce such bright blood images. If it is desired to bring out the images of the arteries or veins selectively, clinicians can take advantage of their tendency to flow in opposite directions and apply a spatially selective preconditioning excitation upstream along the inflow direction of the vessels we wish to suppress, to reduce the magnetization of (saturate) the corresponding inflowing blood in the imaging plane.

A different time-of-flight effect may be seen when detecting the signal as a spin echo. The initially excited spins must also experience the refocusing pulse, which is typically applied in a spatially selective way, to produce a detected signal. If some of the initially selectively excited spins move out of the imaging excitation region before the application of the refocusing pulse, the final resulting spin echo is reduced in intensity. In regions of slower flow, there may not be much intensity loss; this may make it difficult to distinguish slow flow from thrombosis in the images.

Phase shift effects have a somewhat different origin. In conventional imaging, regional phase differences that may develop as a result of local differences in density (e.g., aerated lung is much darker than most other tissues). Most tissues, however, have similar density. Thus, clinicians instead rely on relaxation time differences for most useful image contrast. The delay between initial excitation and signal detection (the echo time, TE) allows some decay of the transverse magnetization to take place. For a given value of TE, the shorter the local value of T2, the greater the amount of this signal decay and the darker the region will appear in the final image. Imaging with a longer value of TE results in a greater relative amount of this T2-weighting effect, in which regions with longer T2 tend to appear relatively brighter.

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of the gradients used in selective excitation and frequency encoding are designed to be balanced out by compensating gradient pulses, so that the phases of stationary spins are uniform at the time of signal detection as a gradient echo (except for any phase encoding gradient effects). If there is motion of excited spins during the image acquisition sequence, however, a spin that was initially found in one region during application of a gradient pulse may have moved to a different region by the time of the application of the corresponding compensating gradient and thus will experience a different local field. Thus, the spin may be left with a net phase offset relative to stationary tissue (Fig. 8). The higher the velocity, the further it will have moved between the two gradient pulses and the greater the phase difference will be. In regions of disordered flow, for example, in a turbulent jet associated with a stenotic valve, the resulting variations in phase may result in corresponding local signal loss that can help delineate the presence and extent of the jet in the images.

One can take advantage of the motion-induced phase shifts specifically to design MR imaging techniques to measure flow velocity, through the phase changes resulting from motion between a pair of extra gradient pulses added to the imaging pulse sequence and designed to sensitize to motion along a particular desired direction. In general, the possibility of local phase variation owing to other causes makes it necessary also to acquire a reference image without the velocity sensitizing gradient to correct for baseline phase variations.

These motion effects can be applied in MR imaging to the study of regional myocardial motion as well as to blood flow. In one approach, myocardial velocities with phase shift imaging methods can be measured, as described earlier. Alternatively, preconditioning phases can be used to alter the local magnetization and noninvasively create a pattern of MR imaging-visible tags within the myocardium (e.g., with spatial modulation of magnetization, SPAMM); motion of the heart during the interval between the tagging and imaging results in a direct visualization of the regional motion and deformation of the heart wall through the corresponding motion and deformation of the tagging pattern.

GATING

Motion of the heart and pulsation of the blood during signal acquisition can result in degradation of the images of the heart and blood vessels. To avoid this, the imaging acquisition can be synchronized with the cardiac cycle, so that image data from a given region are acquired at a consistent phase of the cardiac cycle. This is generally done through monitoring the electrocardiogram (ECG) and triggering the data acquisition from the detected QRS complex. A potential problem can arise through spurious peaks that may be seen in the ECG tracing when the subject is in the magnet, owing to voltages induced by the flow of (electrically conducting) blood across the strong magnetic field of the MR imaging system. In such a case, it may be necessary to reposition the ECG sensing electrodes to a location where these flow-induced voltages are smaller. Although a peripheral pulse sensor may be used to detect the cardiac cycle, the delay between cardiac contraction and the peripheral pulse and the frequent presence of beat-to-beat variability make this generally a less desirable approach for cardiovascular imaging.

In cardiac gated spin echo imaging, the detected QRS complex can be used to trigger data acquisition (with a given value of the phase encoding gradient) from a desired slice location. Because this typically takes only a small fraction of the cardiac cycle, imaging data can then be acquired from other locations, at progressively greater delays after the QRS trigger, until the end of the cardiac cycle.

Figure 8. Phase shift effect. A pair of balanced pulses of gradient (G) results in compensation and no final net phase shift \( \Delta \phi \), for stationary spins \( v = 0 \). Moving spins \( v > 0 \), however, are not fully compensated and are left with a net phase shift dependent on their velocity.
The process can then be repeated with different values for the phase encoding gradient. The net result is a set of images of different locations, each acquired at a well-defined but different phase of the cardiac cycle with prospective gating.

In gradient echo imaging with short TR, an alternative option is available, to use retrospective gating. In this case, the QRS trigger is monitored, but imaging data are acquired continuously, asynchronously with the cardiac cycle. The timing of each such data acquisition within the cardiac cycle is recorded, and the phase encoding gradient value is incremented with each new cycle. At the time of image reconstruction, the acquired data sets are interpolated to create a new synthetic data set, corresponding to specific times in the cardiac cycle. The final result is a set of images at one location at multiple times in the cardiac cycle. As a variation, image data acquisition can move back and forth between different slice locations on successive excitations, resulting in images of more locations but with correspondingly poorer temporal resolution.

With image data acquisition as described previously, only one phase encoded value data set is acquired (for each cardiac cycle phase to be imaged) per heart beat. Because the final image resolution is limited by the number of different phase encoded value data sets acquired, this limits the minimal duration of data acquisition. For example, if it is desired to acquire 128 phase encoding steps (to reconstruct an image matrix with 128 lines along the corresponding direction), the clinician needs to acquire data over 128 heart beats. This necessity precludes imaging in suspended respiration. To get around this limitation, if TR is sufficiently short, the clinician can acquire groups of different phase encoding steps in each heart cycle and treat them as if they were all acquired at the same time. This grouping of multiple phase encoding steps in each heart beat results in a corresponding decrease in temporal resolution but can lead to a shortening of the time required for image acquisition sufficient to permit imaging in suspended respiration.

An alternative approach to shortening the imaging time is to acquire multiple echoes after each excitation and apply a different phase encoding value for each echo (again, treating them as if they were acquired at the same time). These echoes can be acquired as a combination of gradient echoes and spin echoes. In the limiting form of this approach (echo planar imaging), the full data set needed for an image is acquired after just one initial excitation.

TECHNICAL ASPECTS OF MAGNETIC RESONANCE IMAGING

Magnets

The magnetic fields needed for imaging are strong. Typically, they are generated with a magnet using superconducting wires. These are made of a special alloy that has no electrical resistance when cooled to temperatures on the order of liquid helium. This minimizes the need for electrical power to maintain the magnetic field but does lead to costs associated with the need to maintain the necessary low temperatures. Superconducting magnets are the only practical way to achieve high magnetic field strengths.

Magnetic fields themselves, at the strengths found in conventional MR imaging systems, are not harmful. They can exert powerful attractive or torquing forces on ferromagnetic objects, however, posing a potential safety hazard for patients with metallic foreign bodies (particularly in the orbits). Certain intracranial aneurysm chips are ferromagnetic and may be twisted in the magnetic field. Most metallic surgical devices (including all prosthetic heart valves after the early Starr-Edwards models), however, do not pose a significant risk in the magnetic fields of MR imaging systems. A greater potential risk is posed by ferromagnetic objects that may be inadvertently brought into the immediate vicinity of the magnet, where they can become dangerous missiles and damage the equipment and any people who get in their way. Another potential hazard of the magnetic field is in patients with cardiac pacemakers, which may get set to a baseline level in the magnet; if this is not appropriate for the patients, there could be adverse consequences.

To minimize magnetic interference with the environment, the magnetic field around the magnet can be reduced with magnetic shielding in the walls of the room enclosing it. This creates an additional cost in constructing the site for the system but reduces the potential "real estate" costs.

The larger the physical size of the bore space for a patient in a given strength magnet, the more expensive it is to construct. A small
space for the patient can bring out feelings of claustrophobia.

Radiofrequency Fields

The RF fields generated in the MR imaging system can be powerful. This can cause a potential hazard if electrical leads (e.g., to monitor the ECG) are inadvertently arranged so that an effective loop antenna is formed, in which significant current can be induced; this can lead to patient burns. To avoid this situation, the ECG leads are brought straight out of the magnet bore without any loops. An effective loop of this sort can even be set up in the conducting lead in a Swan-Ganz pulmonary artery line or between this lead and the ECG leads, so imaging with such catheters in place should be avoided.

The nuclear MR signal is weak and is easily swamped by noise in the environment. To avoid this, the MR imaging system is typically housed in an electrically isolated environment. This environment can be achieved by creating a Faraday cage with electrically conducting material completely surrounding the imaging system.

Receiver Coils

For conventional cardiovascular imaging, the body-sized receiver coil built into the MR imaging system generally provides adequate SNR. For higher resolution or faster cardiac imaging, however, it may be necessary to use tailored receiver coils designed to receive signal selectively from the region of the heart to improve the SNR sufficiently. This decreases the imaging field of view, so that the coil positioning may need to be adjusted for optimal placement to cover best the heart.

Magnetic Field Gradients

The resolution achievable in MR imaging is in part limited by the strength of the magnetic field gradients used for position encoding. The larger the size of the gradient-generating coils, the more power is necessary to generate a given strength gradient (and the more cost is associated with the system), again limiting the size of the patient imaging space.

To achieve rapid imaging as described earlier, clinicians must be able to switch the gradients on and off rapidly. A physiologic limitation to this switching speed may result from the possibility of producing peripheral nerve stimulation with sufficiently rapid changes in local magnetic field.

Computers

A factor in the rapid development of MR imaging has been the ongoing rapid advances in the capabilities of relatively inexpensive computers. Computers are involved in many aspects of MR imaging, from the control of the different components of the imaging system and reconstruction of the images to the management and display of the resulting image sets, and continuing improvements in their capabilities should further the continued development of MR imaging.

SUMMARY

MR imaging has made rapid progress and promises to be of great utility in the evaluation of the cardiovascular system. Some of the features that make it so promising are its safety (with appropriate guidelines); its ability to produce high-quality tomographic images in arbitrary orientations; and the possibility to obtain unique data, such as on regional myocardial function and metabolism. Ongoing technical developments in such areas as more rapid imaging and newer contrast agents should continue to increase the usefulness of cardiovascular MR imaging.

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