Basic Principles of Magnetic Resonance Imaging

Joseph C. McGowan, PhD, PE[a,b,*]

KEYWORDS

- MR imaging • MR physics
- Magnetic resonance • Spin echo • Gradient echo
- K-space • Fast spin echo

Magnetic resonance (MR) imaging has become the dominant clinical imaging modality with widespread, primarily noninvasive, applicability throughout the body and across many disease processes. This progress of MR imaging has been rapid compared with other imaging technologies and it can be attributed in part to physics and in part to the timing of the development of MR imaging, which corresponded to an important period of advances in computing technology. The history of MR imaging dates from the experiments of Paul Lauterbur[1] in the 1970s and includes the work that resulted in his being awarded the Nobel Prize. Compared to radiography, which develops contrast based upon density, MR imaging probes at least three fundamental parameters and a number of derivative ones. For this reason, exquisite soft tissue differentiation is possible. The flexibility of MR imaging enables the development of purpose-built optimized applications. Concurrent developments in digital image processing, microprocessor power, storage, and computer-aided design have spurred and enabled further growth in capability. Although MR imaging may be viewed as "mature" in some respects, the field is rich with new proposals and applications that hold great promise for future research health care uses.

MR imaging is a modality that richly rewards the clinical practitioner who understands the underlying physics. Compared to other imaging techniques, there are many degrees of freedom in the acquisition parameters for MR imaging. There are opportunities for standardization and also opportunities to tailor protocols to specific disease manifestations and even to specific patients. A variety of artifacts can potentially compromise image quality and their presence can be mitigated or addressed with choices of technique. In MR imaging, there is always an opportunity to trade speed for quality (or vice versa), and the balance point of this compromise may yield an optimum study that maximizes benefit to the patient. The interested reader is encouraged to investigate these topics further in any of several more comprehensive texts.[2–4]

SPIN PHYSICS

When the physics of MR imaging is discussed in the classical sense, the fundamental concept is that of "spin" or of "a spin." Spin refers to a magnetic moment that results from or is associated with a "current loop" created by a spinning charged particle, where the charge resides on the outer surface of the particle. This current can be quantified as

$$I = q * v/2\pi r$$

with *q* the charge on the particle, *r* the radius of the particle, and *v* the tangential velocity of a point on the surface of the particle. For clinical MR imaging, the spin of interest is most often that associated with a proton of water.

The magnetic dipole moment that results is the product of the area of the particle and the current. It is a vector quantity and has direction that is

[a] Drexel University, School of Biomedical Engineering, 3141 Chestnut Street, Philadelphia, PA 19104, USA
[b] Exponent, Inc., 3401 Market Street, Suite 300, Philadelphia, PA 19104, USA
* Drexel University, School of Biomedical Engineering, 3141 Chestnut Street, Philadelphia, PA 19104.
E-mail address: jmcgowan@exponent.com

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parallel to the angular momentum of the spinning particle. The equation

$$\mu = q/2m * \mathbf{J}$$

quantifies the dipole moment with J the angular momentum, q the charge and m the mass of the particle.

Spins tend to align with an external magnetic field, in the same way that iron filings align with a magnetic field in free space. There are two preferred alignments for spins, referred to as "up" and "down" relative to the direction of the applied field. A slight energy difference exists between the two orientations, and thus a system of many spins can be characterized by its energy state. Over time, when spins experience an external field, they will align to a lower energy, or equilibrium, state, characterized by the distribution between the up and down states. A stronger field will develop stronger polarization between the two states, and when the external field changes for a given system of spins a process of "relaxation" to the new equilibrium state will transpire. When a large number of spins are considered as a system, the effect described is observed as that of a single magnetic moment with a direction that can vary in three dimensions. The energy transfer processes that underlie this effect are exploited by MR imaging techniques to gain information about the relevant spins.

When spins are "polarized" or aligned by a static external magnetic field, they can be excited as an ensemble by the application of radiofrequency energy and, given a large enough number of spins, the effects can be detected. The size of the ensemble required for detection is also related to the strength of the external field- giving rise to some fundamental tenets of MR imaging: a larger sample gives a better signal, and a larger (stronger) magnet will do the same, all else equal.

The excitation of the spin ensemble is achieved via resonance, which is the familiar phenomenon by which certain systems can be most effectively excited at a certain frequency. A common example of this is experienced when pushing a child on a swing. If every sequential push is performed with a constant force there is a certain frequency of pushing that will result in the largest effective energy transfer and consequently a higher arc for the child/swing system. Pushing faster or slower has a diminished effect. In magnetic resonance, the characteristic frequency depends upon the characteristics of the spin under investigation and the strength of the applied magnetic field as:

$$f = \gamma B$$

where gamma is the gyromagnetic ratio, a fundamental constant for a given spin, and B the field strength. This famous relationship is known as the Larmor equation. Excitation of the spin system through application of energy at the resonance frequency can be shown to be equivalent to exposing the system to an altered magnetic field, eg, one that is perpendicular to the applied external field. The spins respond, as given above, by relaxing toward the (new) equilibrium state that is associated with the altered field. Thus they relax or move toward an axis that is not parallel to the (large) applied static field.

The motion of the vector representation of a spin ensemble toward an equilibrium position is essential to understanding the basis of MR imaging. That motion is precession, analogous to that observed in a gyroscope (Fig. 1). Precession is characterized by the path taken by a spinning object under the influence of an external force, eg, in the case of a gyroscope: gravity. A spinning gyroscope does not behave as it would were it stationary. Instead of falling over in response to gravity, it "spins down" toward a final rest position at a rate related to its mass and spin velocity. Similarly, when a spin system, aligned with the main magnetic field, is exposed through resonance radiofrequency irradiation to an effectively altered field, it will pass from rotation about the axis of the external field to precession around the axis of the effective field. Upon removal of the resonance excitation, the spin system will again experience only the external field and will precess toward its former equilibrium position.

The motions of spins, or, more correctly, ensembles of spins, were elucidated by Bloch and Purcell who for this work shared the Nobel Prize in 1952. Bloch[5] described this motion with a series of coupled differential equations (Bloch equations) that when solved give rise to two exponential relaxation constants known as T1 and T2. The T1 constant is associated with spin–lattice relaxation and describes the exchange of energy between the spin system and the environment. T1 also (or equivalently) describes the longitudinal relaxation along the z-axis. The state of z-magnetization can be normalized to between 1 (equilibrium) and −1. The T2 constant describes the coherence of the spin system, a measure of how "together" the spins are as they rotate. This can be normalized to 1 for the situation where all of the spins are aligned, and to zero when describing a completely random orientation between spins within the system.

A special case of motion of the effective spin vector is seen when the vector rotates perpendicular to the main magnetic field. If the main field is assumed to be oriented in the z-direction as is conventional, this spin rotation is said to be in the x–y plane. With the spins rotating in this

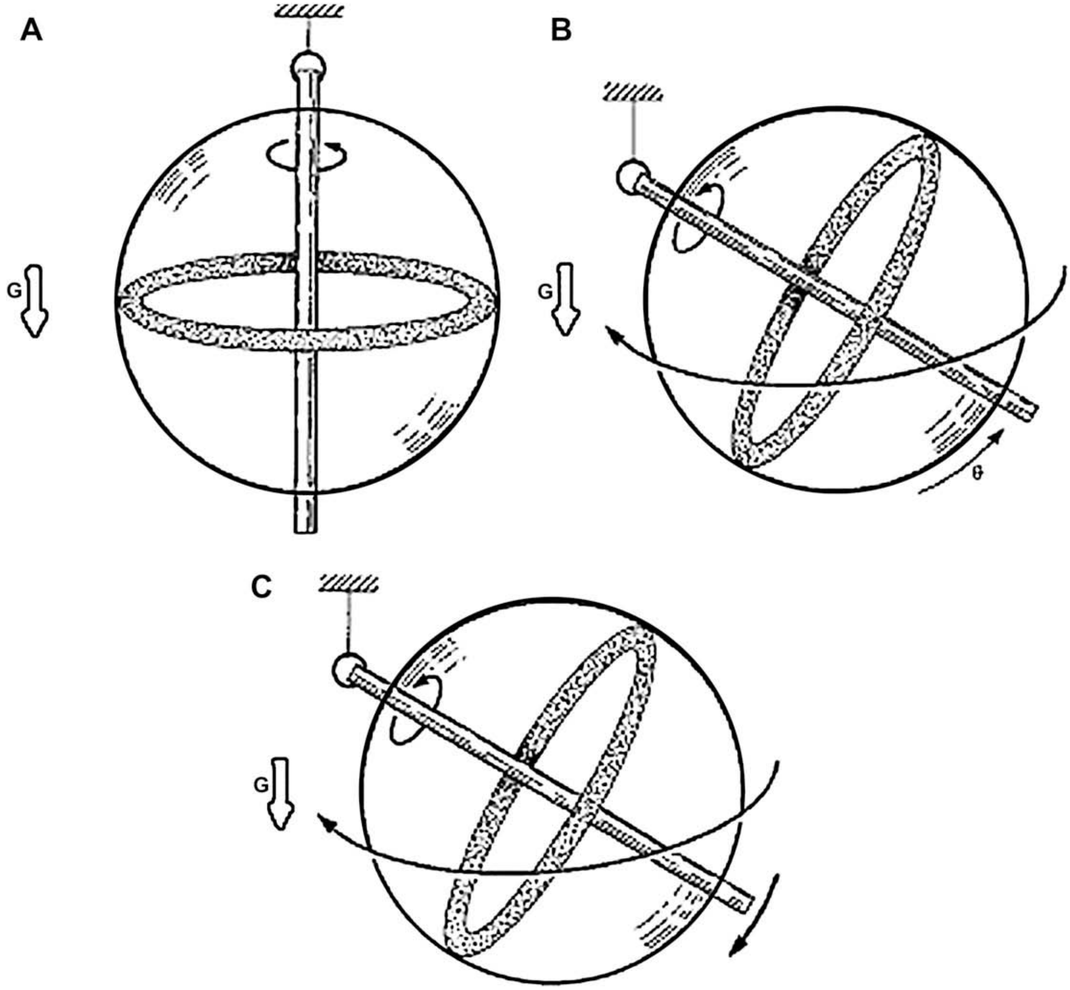

Fig. 1. (*A*) A spinning gyroscope is suspended in alignment with the gravitational field. If the gyroscope is displaced by the application of a force, it will continue to rotate about its physical axis but will also precess about the direction of gravity (*B*). As the rotational velocity of the gyroscope decreases, the angle θ approaches zero (*C*). (*From* McGowan JC. Magnetic Resonance. In: Brans, Hay, editors. Physiological monitoring and instrument diagnosis in perinatal and neonatal medicine. New York: Cambridge University Press; 1995. p. 67; with permission, Cambridge University Press.)

manner, it is possible to introduce a coil of wire such that the spins effectively serve as magnets whose lines of flux cut the coil, giving rise to Faraday induction and an induced voltage in the coil. This principle is exploited to obtain the signal associated with the spin behavior at a given time, resulting in data from which an image can be reconstructed.

MR IMAGING HARDWARE

Principal components that comprise an MR imaging system include: a main magnet, gradient magnets, radiofrequency coils to transmit and/or receive the MR imaging signal, and associated signal processing equipment.

State of the art clinical MR imaging systems typically incorporate a superconducting magnet to produce a relatively large static magnetic field. The function of this magnet is to produce a stable, spatially homogeneous magnetic field over a prescribed volume. Such a magnet may operate at 1.5 Tesla (T), or 15,000 Gauss, which can be compared with the earth's magnetic field at approximately 0.5 Gauss. Modern magnet systems can also be obtained at 3T and higher fields, and there are also lower-field magnets available, typically at lower cost or for special purposes. Superconducting magnets are cooled by cryogens such as liquid nitrogen and liquid helium and, after the static field is established, they require no electric power to maintain it. Cost, weight, and siting are often important considerations when installing a magnet system. Increased field strength is desirable from the standpoint of signal to noise ratio (SNR) (ie, signal quality) but carries the disadvantages of increased cost largely because of the need for more wire to carry the high current as well as the increased size of the cryostat. In addition, fringe fields must be designed for and the footprint of the device may be large relative to other imaging equipment.

The main magnet will typically incorporate the capability of making small adjustments to the field to achieve the homogeneity necessary for MR imaging. This may be accomplished with secondary magnetic fields known as "shims," which add to the main field in a spatially inhomogeneous way to correct errors and compensate for engineering factors.

Another kind of secondary magnetic field incorporated for MR imaging is known as a gradient. Gradient coils are electromagnets installed so that when energized, they predictably perturb the main magnetic field along an axis. Thus, in the z-direction, the effective (net) field in the presence of a gradient can be given as

$$B_{net} = B_0 + \Delta B_0 = B_0 + G_Z Z$$

where B_0 is the strength of the main magnetic field, ΔB_0 is the contribution to the effective field from the gradient, G_z is the strength of the gradient in the z-direction, and Z is the distance along the z-axis (**Fig. 2**). The value G is usually given in mT/m. Referring to the Larmor equation given above, it is seen that in the presence of a gradient, the resonance frequency associated with the sensitive volume of the main magnet varies along the gradient axis. Assuming that the variation of the field along that axis is linear, the frequency variation will also be linear. This observation will form the basis for the MR imaging techniques discussed below.

To obtain three-dimensional images, it is required that at least three gradients be present. An additional practical requirement is that they must be able to be switched on and off very rapidly. The isocenter of the three gradients should be co-located with the isocenter of the main magnet.

Radiofrequency coils are used to excite a particular volume of interest by adding energy to the tissue or material within that volume. They also serve to detect the eventual MR imaging signal. In some applications, both transmit and receive functions are performed by the same coil; in other systems, it is advantageous to employ separate transmit and receive coils. Analogous to the swing example given above, the radiofrequency coils are tuned to the resonance frequency of the spins and thus

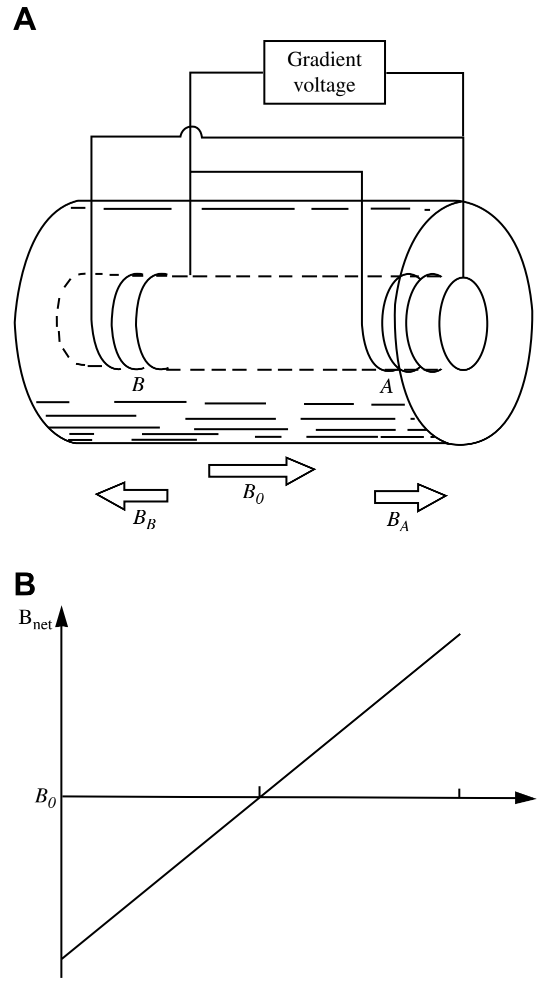

Fig. 2. (*A*) Application of gradient voltage produces gradient magnetic fields B_A and B_B. The effective magnetic field B_{net} is obtained by adding, at each location along the z-axis, the contribution from all three fields. As shown in the graph (*B*) B_{net} increases with increasing z and is equal to B_0 at the magnet center. The effective combination of B_a and B_b is equal to the ΔB_0 term given in the text. (*From* McGowan JC. Magnetic Resonance. In: Brans, Hay, editors. Physiological monitoring and instrument diagnosis in perinatal and neonatal medicine. New York: Cambridge University Press; 1995. p. 75; with permission, Cambridge University Press.)

they are able to add energy in an effective manner to the spin systems. Radiofrequency coils can be constructed with a variety of geometries and they are increasingly built for specific applications. They vary from the large "body coil" that is supplied with many magnet systems and still used for imaging of larger portions of the body, to coils designed for a single finger or perhaps an internal organ. There is a substantial advantage to tailoring the size of the coil to the tissue being examined because the signal in the image must only come from the tissue of interest, whereas the noise associated with that image will originate in the entirety of the coil. Because the image quality is related to the SNR, it is apparent that the highest quality will in general be achieved with coil size optimized to the tissue of interest. It is not only possible but also ubiquitous to "trade" speed for quality in MR imaging. Whenever sufficient quality exists in the acquisition, one can opt to image faster while sacrificing a predictable amount of quality, measured as the SNR.

Radiofrequency coils are increasingly provided as arrays of individual coils that are used sequentially during the imaging sequence. This design can be used to increase quality by redundant or semiredundant acquisition of data, or to increase coverage of an area by establishing a coil array with a predominant linear dimension, for example, to image the spine. This is a way of increasing SNR, which can be used to produce images with higher quality or, alternatively, to increase the speed of acquisition.

When voltage is induced in the radiofrequency coil, the voltage is amplified by a receiver and sent through a signal processing train to be stored in a computer. The computer will carry out the reconstruction of an image when sufficient data is obtained. When phased arrays of coils are used, each must be connected to a separate channel for signal processing.

RELAXATION AND THE MAGNETIC RESONANCE SIGNAL

The state of a spin system can be described in terms of the amount of longitudinal magnetization (z-direction) and the amount of transverse magnetization (x–y direction). The equilibrium position of the magnetization is fully aligned with the z-direction. When energy is added to the spins, the magnetization is "tipped" or "flipped" such that there can be a component of transverse, or x–y magnetization. The largest amount of transverse magnetization is achieved by tipping the spins ninety degrees. The state of the spin systems can be described by or decomposed into two vectors, one constantly pointing along the z-axis and one representing the projection of the total magnetization in the x–y plane. A spin system that is not at equilibrium energy will tend to relax to equilibrium as predicted by the Bloch equations. This relaxation is governed by the two exponential relaxation constants T1 and T2. T1 relaxation, also known as spin–lattice relaxation, describes the motion of the z-magnetization vector as it proceeds to its equilibrium state. To understand T2 relaxation, one must recognize that when the spins are tipped into the transverse plane, they can get out of phase with one another because of the presence of small variations in resonance frequency arising from differences in the spatial environment, chemical environment, or random processes. As the spins in the transverse plane tend to be less aligned, they begin to cancel each other out so that the net transverse vector gets progressively smaller. The time constant for this process is T2 and this is called spin–spin relaxation. The composite relaxation process can be envisioned as precession, which is analogous to a gyroscope relaxing to an equilibrium state in the presence of gravity.

The presence of transverse magnetization (and only transverse magnetization) induces voltage in the radiofrequency coil. This is illustrated by a simple coil design whereby a loop of wire is placed within the static magnetic field and aligned as shown in **Fig. 3**. Faraday's Law provides the requirements for inducing a voltage: a conductor, a magnetic field, and relative motion between the two. These requirements are met by the coil (serving as the conductor), spin system (as the magnetic field), and the motion of the transverse magnetization serving to make the lines of flux associated with the spin magnetization cut through the conductor of the coil (**Fig. 4**).

Assuming that energy has been added to a spin system via excitation at the resonance frequency, a coil in position to detect a signal will have induced in it a decaying sinusoidal voltage that is known as a "free induction decay" (FID). Note that a FID begins immediately after the excitation

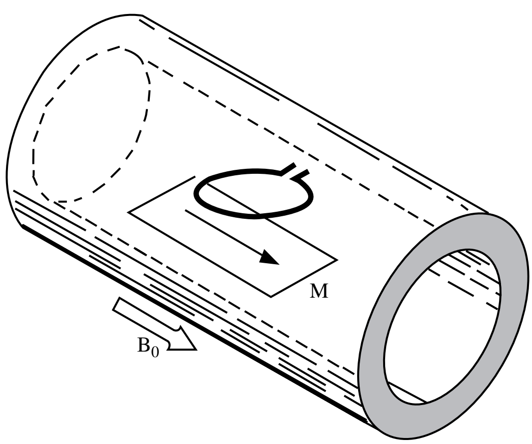

Fig. 3. The sample magnetization M is aligned with B_0 and in proximity to a conducting coil. (*From* McGowan JC. Magnetic Resonance. In: Brans, Hay, editors. Physiological monitoring and instrument diagnosis in perinatal and neonatal medicine. New York: Cambridge University Press; 1995. p. 69; with permission, Cambridge University Press.)

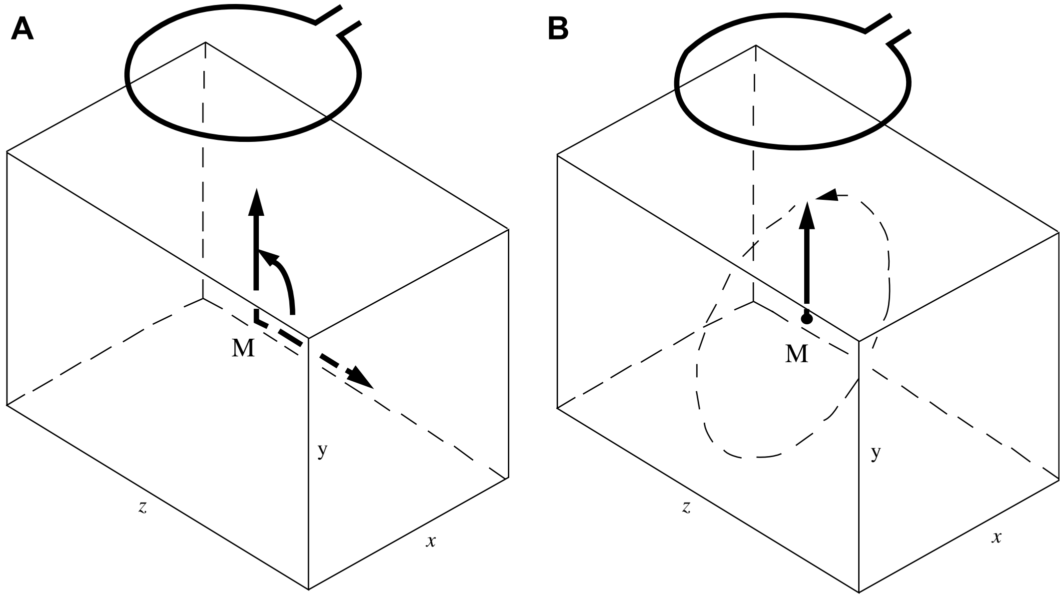

Fig. 4. After the magnetization M is tipped from the z-direction by ninety degrees (*A*) it begins to precess in the x–y plane (*B*) and satisfies the requirements for Faraday induction. (*From* McGowan JC. Magnetic Resonance. In: Brans, Hay, editors. Physiological monitoring and instrument diagnosis in perinatal and neonatal medicine. New York: Cambridge University Press; 1995. p. 69; with permission, Cambridge University Press.)

ceases. This signal can be useful, particularly in the area of spectroscopy. However, to perform imaging it is often desirable to separate the signal in time from the excitation pulse. To do this, one can manipulate the phase state of the spins in the transverse plane. As stated above, the spins dephase through a combination of random (or irreversible) and nonrandom (potentially reversible) processes; dephasing is directly related to the magnitude of signal. The spins can be caused to dephase (and subsequently rephase) by the application of a gradient along one axis. Recalling the Larmor relationship, the spins experience a decrease or increase in frequency that is proportional to the change in applied local field compared with the static field. Thus, the spins begin to get out of alignment with one another and they are said to acquire a phase difference or to dephase. To rephase the spins, the gradient amplitude is at some point in the sequence simply reversed. Immediately, the frequency differences between spins are reversed and the spins begin to acquire phase in the opposite direction. As the spins are rephrased, the signal will reflect a buildup to a maximum magnitude followed by a decay in magnitude, and this “gradient echo” can be positioned as desired, in time, through the timing of the gradient application.

A different dephasing/rephasing process results from the small inhomogeneities present in the static field. These are not aligned with an axis like the variations a field gradient would impose, but they are spatially invariant. This dephasing, which occurs naturally following excitation, is most effectively reversed via the application of a radiofrequency pulse corresponding to a 180-degree flip angle, and the resulting signal is called a spin echo (or Hahn echo, after Erwin Hahn, the discoverer of the effect).[6] The spin echo signal is displaced from the initial excitation by a time called “TE” (echo time) (with the second RF pulse at time TE/2) and also builds up and decays in a similar manner to that described above. Two depictions of the effects of rephasing the reversible dephasing are given in Fig. 5. In these examples, one can see the effects of the inversion pulse on a simple two-spin system with the phase position tracked in time (see Fig. 5), and, equivalently, in the time-domain representation of sinusoids with different frequencies corresponding to several spins (see Fig. 5). In a system with many spins, the effect is more dramatic because the signal essentially vanishes for a period of time, and then reappears at the predicted time of the spin echo.

GRADIENT ECHO IMAGING

Through the application of field gradients, it is possible to change the magnetic field of each point inside the MR imaging machine. Similarly, it is possible to establish a single point in the magnet that has a particular resonance frequency through

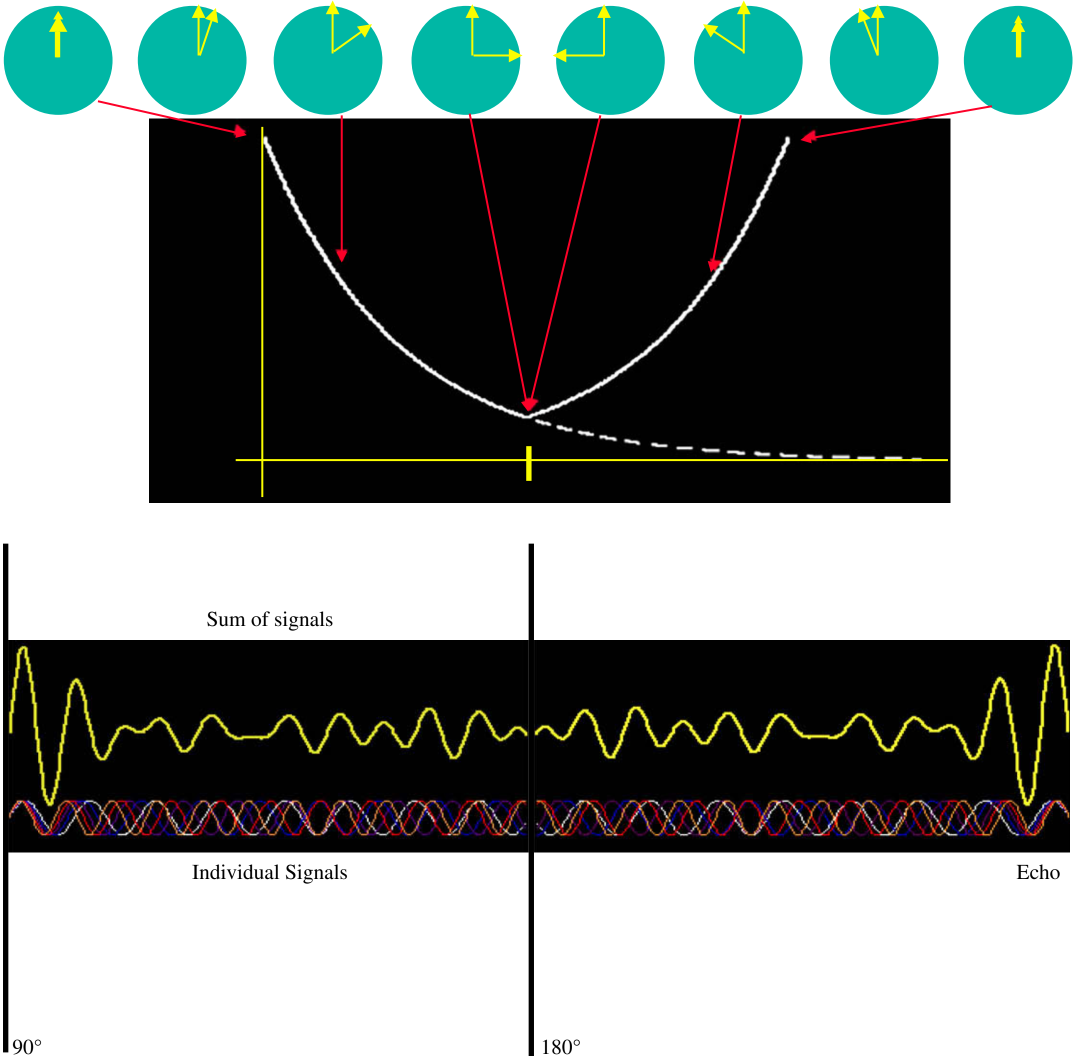

Fig. 5. A two-spin system is depicted following excitation whereby the spins are tipped into the transverse plane. The signal, consisting of the addition of the two spins, is maximized when the spins are in phase. As one spin gains phase relative to the other, the total signal decreases (left to right on the figure) until a 180-degree pulse is given (yellow tick on the signal graph) and the phase difference now tends to rephase the spins (top). When the signals are viewed as sinusoids in the time domain, their individual magnitudes are constant but their sum varies with the state of phase. Three sinusoids are shown to dephase, experience a 180-degree pulse, and rephase to an echo.

manipulation of the gradients in all three dimensions. If excitation is performed at that frequency, the information gathered from the single point in space can be stored and later combined with others to form an image. This method of imaging was dubbed the "sensitive point" method and represented an early attempt to exploit MR imaging.[7] However, this method is impractical for clinical imaging because of the time required for data acquisition. Modern imaging employs the field gradients to encode spatial information and, using the Fourier transform, allow the simultaneous acquisition of data from many points along an axis.

The first step in the formation of a 2D magnetic resonance image is to "select" a slice of the tissue under examination. This is done by establishing a gradient in one direction, for example, the z-direction, and applying a radiofrequency pulse that will add energy to the spins. However, the frequency of this pulse is adjusted such that the only spins that have a corresponding frequency (and thus will experience the resonance effect) are those that lie within the "selected" slice by being in the correct position along the z-axis (**Fig. 6**).

Recall the general premise that the application of a field gradient along an axis establishes

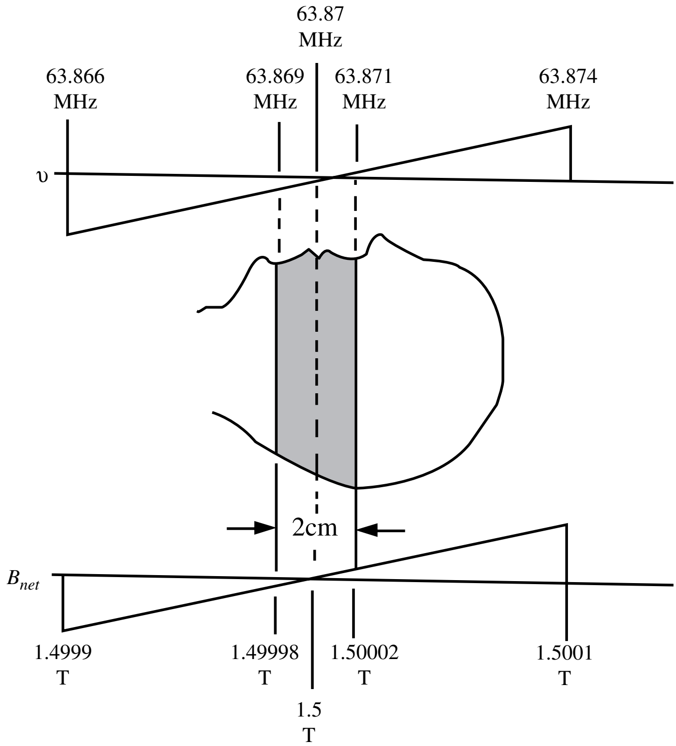

Fig. 6. Application of rf excitation over a range of frequencies in conjunction with a field gradient is used for slice selection. The example depicts a 2-kHz-wide excitation bandwidth, applied with a gradient that changes the B_{net} field by 0.2 G/cm. The rf excitation corresponds to the resonance frequencies of protons experiencing fields of 1.49998–1.50002 T, or a range of 0.4 G. This results in excitation of a slice 2 cm thick that is centered on the magnet center. (*From* McGowan JC. Magnetic Resonance. In: Brans, Hay, editors. Physiological monitoring and instrument diagnosis in perinatal and neonatal medicine. New York: Cambridge University Press; 1995. p. 76; with permission, Cambridge University Press.)

a correspondence between frequency and spatial position. If the signal in the coil is recorded during the gradient application, the spins along the gradient axis will effectively rotate with characteristic frequencies that reflect their position. In fact, the gradient application in one direction, for example, the x-direction, is applied such that it encompasses in time the gradient echo. Thus, the signal in the coil will consist of an effective summation of many signals at different frequencies. The "encoded" information is recovered via the mathematical process of Fourier transformation.

The process of forming and recovering such a gradient echo can be repeated after the application of an orthogonal gradient that establishes a phase relationship along the remaining axis, in the previous example, the y-direction. The mathematics involved in the reconstruction of this data is a two-dimensional Fourier transformation, the details of which are beyond the scope of this article. Applying both phase encoding in the y-direction before the acquisition of signal and frequency encoding in the x-direction during the signal acquisition makes each point in the image plane have a particular and unique history regarding the resonance frequency during the precession of the spins.

The gradient echo pulse sequence diagram is presented as **Fig. 7**. Gradient echo imaging is more demanding with respect to the scanner than spin–echo techniques, because the dephasing that results from inhomogeneities within the static field (but not because of field gradients that are under the operator's control) is not rephased and thus, any signal that is lost on that time scale is not recovered. However, improvements in magnet design and construction have made it possible to acquire good quality gradient echo images and, in fact, at higher fields where spin echo inversion pulses are limited because of power deposition, gradient echo imaging can be the best option.

SPIN ECHO IMAGING

Spin echo imaging has been the workhorse pulse sequence during the history of MR imaging. One reason for this is the excellent signal quality that is achieved when the reversible dephasing is completely rephased. Specifically, in the spin echo method, all dephasing associated with magnetic field inhomogeneities that are stationary, is rephased at the time of the spin echo. Simultaneously, because of frequency encoding gradients that are applied in a manner identical to that explained above, a gradient echo also occurs. The pulse sequence is almost identical to the gradient echo pulse sequence, with the exception of

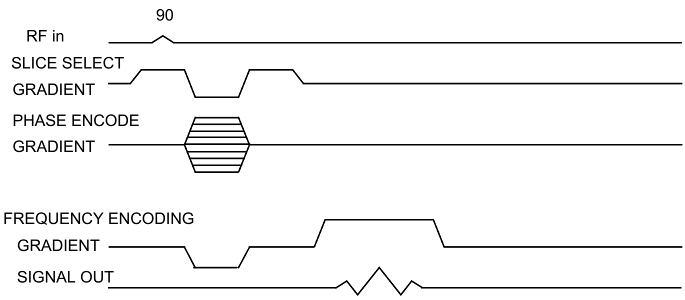

Fig. 7. In a gradient echo pulse sequence a single excitation is applied (90 degrees is shown here) The slice-select gradient is turned on when the rf excitation is applied such that only the desired slice is affected by the rf. The phase encode gradient is turned on between the excitation and inversion pulses, with a different magnitude for each TR. The frequency encode gradient is applied during phase encoding (dephase) and signal acquisition (rephase). As time passes from left to right, when the area under the frequency-encode gradient curve sums to zero all of the dephasing effects of the gradient have been reversed and a gradient echo is formed.

adding the inversion pulse that creates a spin echo at time TE. It is depicted in **Fig. 8**. Slice selection, frequency and phase encoding, and reconstruction of the image are achieved in the same manner as for the gradient echo. Instead of slice selection, yet another phase encoding gradient may be incorporated into the pulse sequence, and the reconstruction of the image performed with a three-dimensional Fourier transform. The result of such a process is three-dimensional MR imaging and it offers advantages that include the ability to obtain smaller slices.

MR IMAGING CONTRAST

Referring to the spin–echo pulse sequence (see **Fig. 8**), one can see that there are two principal timing variables: repetition time (TR) and TE. The TR can be thought of as the time spacing between phase encode pulses, or alternatively and equivalently, as the time to play out one complete pulse train. The TE represents the time between the initial RF excitation and the acquisition of the signal. It has been noted that the spin echo methodology rephases spins that were dephased by spatially invariant magnetic field inhomogeneities. There are other mechanisms of dephasing that are random in nature and thus cannot be reversed. The irreversible dephasing leads to a loss of signal that is described by the T2 time constant. The reversible dephasing that is refocused by the spin echo can be combined with the T2 processes and the composite time constant is known as T2*. T2 decay is thus always slower than T2* decay, and usually markedly so. The effects of T2 and T2* can be further understood by considering that the decay of a FID reflects T2*, while the difference in signal strength in spin echo acquisitions with differing TE reflects T2.

The effect of changing TR can be understood by considering two spin systems with equal equilibrium magnetization but differing T1 decay constants. An example in the brain would be the lateral ventricles, more water-like with long T1, and brain parenchyma, with shorter T1. Both tissues can be excited by a 90-degree radiofrequency pulse and allowed to return to equilibrium during a long repetition period (TR) after which the process is repeated. In this example, after each excitation, the magnetization is at its maximum value, and after each excitation, the signals from short and long T1 regions are identical. If

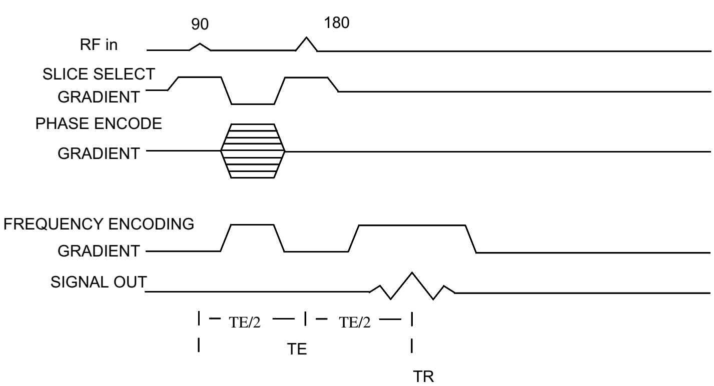

Fig. 8. In a spin–echo pulse sequence the rf excitation can take the form of a 90-degree pulse followed at time TE/2 by a 180-degree inversion pulse. Slice select and phase-encode gradients are applied as discussed above. The first positive lobe of the frequency encode gradient (coincident with the phase-encode gradient) is effectively reversed by the 180 degree pulse and so the point of the gradient echo (where the effects of the frequency encode gradient sum to zero) corresponds to the spin echo at time TE.

the equilibrium magnetizations from the spin systems were not equal, the signal from each would reflect the number of spins or the spin density. In MR imaging, an image reflecting these differences is called a proton-density image and is analogous to an x-ray image. Proton-density images are less often prescribed compared with relaxation-time-weighted images, but there are examples where their use is preferred. On the other hand, if the TR is shortened, the situation can arise where one spin system relaxed back to equilibrium but the other, experiencing longer decay times, did not. In this instance, the subsequent excitation pulse will cause the fully relaxed spin system to attain full transverse magnetization (and signal strength), but the partially relaxed system will attain less than full transverse magnetization. This can be understood by imagining that the short T1 spins are at a longitudinal magnetization of 1, but the long T1 spins, having not fully relaxed, might be at a longitudinal magnetization of 0.5. A subsequent excitation would result in high transverse magnetization for the short T1 spins, but low for the long T1 spins. The system with the longer T1 is said to be partially saturated in this case. If imaging is performed in this way, the spins with longer T1 (for example, those from fluid as opposed to tissue) will appear hypointense on the image. Since the image demonstrates differences in T1, this is called T1-weighted imaging. T1-weighted imaging is often excellent for demonstrating anatomy. Note that although longitudinal magnetization is not detected directly, the maximum transverse magnetization (and thus signal) that can be achieved directly reflects the state of longitudinal magnetization immediately before the excitation pulse. This gives rise to another imaging option, known as inversion recovery.

The longitudinal magnetization of a spin system reaches a maximum value at equilibrium, and the magnitude of this is often called M_0. A composite spin pointing along the z-axis with magnitude M_0 can be rotated through any number of degrees about any axis via the proper excitation radiofrequency pulse. It is possible to rotate that magnetization ninety degrees and, in essence, convert all of the longitudinal magnetization to signal-producing transverse magnetization. In another special case example, the longitudinal magnetization is rotated 45 degrees and at that point the transverse and longitudinal magnetization values are equal. In general, up to ninety degrees, each increase in rotation of the spin vector results in more transverse magnetization and less longitudinal magnetization following the rotation. It is also possible to rotate the spin vector greater than ninety degrees, and as the rotation increases the amount of transverse magnetization now decreases, but the longitudinal magnetization following excitation continues to be farther from equilibrium up to a maximum of a 180-degree excitation. In this final special case, the transverse magnetization following the excitation is zero, and the longitudinal magnetization is M_0. The spin system is said to be inverted. Relaxation proceeds, then, without any transverse magnetization being developed, however, at any time a second excitation pulse can be applied to convert the present state of longitudinal magnetization to transverse and produce a signal. A very interesting and useful case is possible whereby the second excitation pulse is given when the spins associated with a tissue or substance of interest (typically fat or water) pass through the null state where their z-magnetization is equal to zero. When this is done the rest of the tissue (having differing T1) will develop transverse magnetization and produce signal, but the tissue that is going through the null will produce no signal. The image reconstructed from this data will have areas of zero intensity corresponding to the nulled tissue. When the sequence is designed to null water (including cerebrospinal fluid), the technique is called FLuid Attenuated Inversion Recovery (FLAIR).[8]

Returning the TR to a (long) value that minimizes T1-weighting, one can consider the effects of changing the TE. In particular, lengthening the TE allows the signal from some spin systems to completely dephase before it is acquired and thus, upon acquisition, there is no signal. On the other hand, other spin systems (with long T2) will demonstrate substantial signal at the time of acquisition. Thus, the resulting image will demonstrate hyperintensity corresponding to regions of long T2 and hyperintensity corresponding to regions of short T2. This is, of course, T2-weighting.

Summarizing, if TE is short on the T2 time scale of the tissue, a TR that is short on the T1 time scale of the tissue of interest will produce T1-weighting, and a TR that is relatively long will produce proton-density weighting. The combination of a long TR and a relatively long TE is used to obtain T2-weighting. These timing variables can be optimized to develop specific contrast that is relevant to the tissue of interest. In general, the time to acquire an image is equal to the repetition time multiplied by the number of repetitions, equivalent to the number of phase–encode steps. In clinical imaging, it is typical to use 256 repetitions of the sequence, but 512 is not uncommon, and 128 or fewer repetitions is sometimes used in the interest of obtaining a faster examination.

A BRIEF INTRODUCTION TO K-SPACE AND FAST SPIN ECHO IMAGING

Since the first magnetic resonance images were obtained there was interest in acquiring images faster, and with higher quality. During the intervening time, there have been concomitant advances in the technologies of computing and of mechanical design and construction such that current MR imaging scanners offer significantly advanced capabilities. A limitation that was soon observed with the rise of T2-weighted diagnostic scanning was the need for long TR periods, to avoid saturating spin systems and thus adding T1 contrast when "pure" T2 contrast was desired. Additionally, it was observed that the process of T2-weighted imaging was relatively inefficient in that the largest part the time duration of the pulse sequence was spent simply waiting for relaxation to occur. At the same time, it was recognized that the process of refocusing spins with a spin echo technique could be repeated within a single TR period, causing multiple spin echoes to occur within one repetition. If these multiple echo signals are acquired and stored, multiple images can be reconstructed corresponding to different TEs within the same TR. A natural, yet profound, evolution of this technique led to a rapid method for acquiring T2-weighted diagnostic images. Understanding the idea of "k-space" is necessary to appreciate this development and quite helpful in the understanding of many modern MR imaging methods.

Consider that the signal acquired during an MR imaging sequence is recorded over time. Each spin echo (or gradient echo) that is sensed in the receiver coil requires a certain amount of time to play out and the echo is stored in the computer memory, to be combined with the rest of the echoes that comprise the pulse sequence. It was noted that via a Fourier transformation this echo data is converted, or reconstructed, into an image. The Fourier transformation is a mathematical tool that can be thought of as a decomposition of a complicated signal into component parts, each with a characteristic frequency and magnitude. This is a particularly useful tool for MR imaging because, as has been noted, the spatial position information is encoded into the signal via the function of gradient magnetic fields that establish a correspondence between frequency and spatial position. Thus, the signal detected in the receiver coil is a composite of many frequencies. The signal is referred to as a time–domain signal because it is simply a recording of signal strength in time. The Fourier transformation breaks this signal into component frequencies and magnitudes. If the result of this were plotted on a graph, the ordinate could be labeled frequency. Hence, this is a frequency–domain representation. The utility of this application in MR imaging is that frequency is related to space (again, via the gradients) and thus the frequency domain is equivalent to the spatial domain. Thus, for each component part of the signal, the Fourier transformation effectively establishes the source position (along the gradient axis). The number of spins at that location (effectively spin density) is preserved in the magnitude of the signal.

A mathematician would view an image as simply a matrix of numbers, each corresponding to a level of intensity on the scale from black to white. Each intensity number of course corresponds to a position in space. A matrix of data from the receiver coil can also be formed by essentially stacking the intensity values from the coil so that the first spin echo is depicted as a row across the matrix, and each successive spin echo (having a different phase encode gradient applied) forms another row. Each intensity value in this matrix, the k-space matrix, corresponds to a position in time, and represents the signal received by the coil at that time. To complete the theoretic picture, it was noted that the Fourier transformation is used to convert from spatial dimensions to time dimensions. This can be in one dimension, as discussed above, or in two dimensions as is related to image formation, or even in higher dimensions to achieve, for example, three-dimensional imaging. The matrix full of data, the k-space matrix, is acquired in time and then the matrix is transformed into the frequency, or spatial, domain where it can be displayed as an image. The two matrices — image space and k-space — have the same dimensions. Thus, one can see that to construct an image, k-space must be "filled" with data with the same number of points as there will be elements in the final image. There is a counterintuitive aspect of k-space and the Fourier transformation in that the signal acquired at a particular time (that is, at a point in k-space) does not directly transfer to the intensity of a single point in image space. Rather, each point in k-space influences all points in image space. Equivalently, each point in image space is contributed to by all points in k-space. This gives rise to the interesting phenomenon that the central region of k-space is highly influential over the contrast in the entire image, whereas the periphery of k-space determines the edge definition and fine structural quality of the image.

The reason that the center of k-space determines contrast is related to the fact that the signal strength corresponding to the central k-space is the strongest. This can be understood by considering the effect of gradients when turned on. That effect is to change the phase of spins and when spins start out in phase, the effect is to dephase them according to their position along the gradient axis. Considering the pattern of phase during a gradient echo, it can be seen that the spins will be most in phase (with strongest signal) at the center of the echo, which is also the center of k-space in the x-dimension. The center of k-space in the y-dimension corresponds to the zero magnitude of the phase-encode gradient, and thus is also the region of least dephasing.

If the k-space matrix is viewed as a blank slate on which the data points can be written, it is easy to understand the concept of "covering" k-space to acquire enough points to reconstruct an image. Turning on the frequency encoding gradient in the foregoing discussion can be envisioned as moving across k-space from left to right (to fill in one row) while the phase encode gradient has the effect of moving up and down so as to prepare to fill in another row. Depending on the design of the pulse sequence, gradients may be used to move in any direction within the k-space matrix.

Returning to the challenge of rapid acquisition of T2-weighted images, recall that it is possible to create multiple spin echoes within a single TR period by simply applying additional RF excitation pulses within the time period. Thus, there are several spin echoes, for example, four, within each TR, and they are grouped by TE. If the resulting spin echoes are recorded and stored as groups, there will eventually be enough data to construct four images: one corresponding to each TE for which spin echoes are collected (Fig. 9). Instead of filling up four groups of echoes, all of the echoes collected are kept together and used to reconstruct a single image (Fig. 10a). In this example, the matrix for that single image will be filled four times as fast compared with the conventional acquisition (see Fig. 10b). This is the premise of "turbo" or "fast" spin echo, first introduced as rapid acquisition with relaxation enhancement (RARE).[9]

The alert reader will have noted that there is a significant difference in the k-space makeup of a fast spin echo (FSE) image compared with a conventional T2-weighted image as discussed above. Specifically, the FSE image is composed of spin echoes that reflect, in this example, four different values of TE. Because the T2-weighted contrast is a function of the TE, this leads to the question of what the contrast in the FSE image means. The answer reflects the observation above that

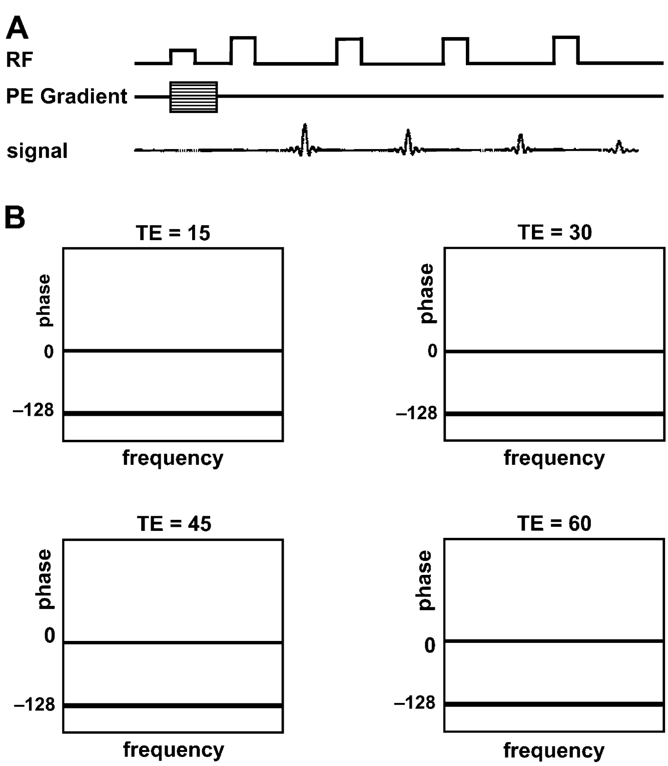

Fig. 9. In a multiple spin–echo sequence, one phase-encode gradient application per TR is used (top). Four images are acquired when the PE gradient is incremented through full range, and the contrast is each image reflects the TE of the echo associated with that image. K-space maps are shown (bottom) reflecting only the first TR period where the PE gradient is at maximum negative amplitude. Each echo signal decreases in magnitude according to T2* and the maximum amplitude decreases with each successive echo due to the effects of T2 decay. (*From* McGowan JC. Fast imaging with an introduction to k-space. In: Filippi, DeStefano, Dousset, et al, editors. MR imaging in white matter diseases of the brain and spinal cord. Berlin: Springer-Verlag; 2005. p. 48; with permission, Springer Science and Business Media.)

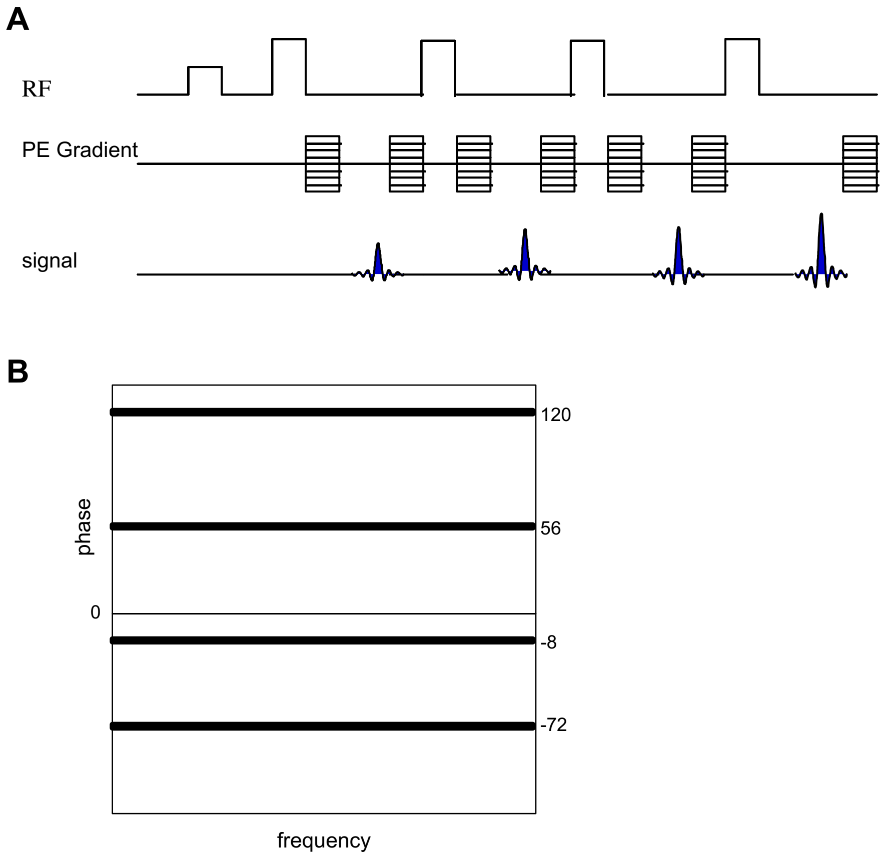

Fig. 10. In a fast spin–echo image, multiple phase-encode gradient applications are used in each TR. One image is acquired after all phase-encode values are applied (*A*). A k-space map for the first TR period is shown (*B*). As depicted, the normalized values of the phase-encode gradients are 120, −72, 56, −8. Ordering in this way makes the signal intensity of the echoes increase during the TR, as the signal gain from reduced dephasing exceeds the signal loss from T2. In this example, if the pattern is continued the contrast of the resultant image will reflect the later echo. (*From* McGowan JC. Fast imaging with an introduction to k-space. In: Filippi, DeStefano, Dousset, et al, editors. MR imaging in white matter diseases of the brain and spinal cord. Berlin: Springer-Verlag; 2005. p. 48; with permission, Springer Science and Business Media.)

the contrast in any image reflects the data in the central region of k-space. Thus, in this example, one can choose to have the central k-space data come exclusively from only one of the TE values. This value will be the effective TE for that image and the contrast in that image will closely resemble the contrast that would be obtained from a conventional acquisition with a single TE equivalent to the effective TE.

The time to acquire a FSE sequence is related to the time given above for a conventional acquisition by the number of spin echoes created and acquired during each repetition time (ie, between excitation pulses). Some manufacturers refer to this number as "echo train length" and others refer to it as the "turbo factor" and indeed the gain in speed is proportional to that number. Thus, the total acquisition time can be found by multiplying the number of phase encode steps by the repetition time and dividing that product by the number of echoes (echo train length, turbo factor) during each repetition.

OTHER RAPID IMAGING TECHNIQUES

One of the first rapid imaging techniques to be introduced followed the advance of magnet technology to the point where usable signal could be obtained without relying on a spin echo to refocus the large amount of reversible dephasing. The Fast Low Angle SHot or FLASH[10] technique exploited the observation that recovery of sufficient longitudinal magnetization to allow a subsequence excitation was much faster if an initial pulse much less than ninety degrees was used. With this imaging technique, it is also not necessary to wait for full relaxation of the longitudinal magnetization as a steady-state condition is achieved and exploited. This gradient echo technique is useful to acquire T1 or proton density contrast. Related techniques are also useful in the clinic as a means of identifying hemorrhage because the dephasing is associated with deposits of blood, a phenomenon that evolves over time.

More recent advances in rapid imaging have included a family of gradient echo techniques known as "echo–planar" imaging (EPI).[11] EPI makes use of very rapidly changing gradients to cover k-space completely following a single excitation pulse. In one variant of EPI, a spin echo is formed with an excitation pulse and a single inversion pulse. During the spin echo, the rapidly oscillating gradients effectively form many gradient echoes that are acquired during the span of milliseconds. EPI finds application whenever subsecond imaging times are required, for example, in the presence of physiologic motion that is too fast to capture with conventional or even FSE imaging. Functional imaging of the brain is also often performed with EPI pulse sequences. Hybrid forms incorporating aspects of both gradient echo imaging and spin echo imaging are increasingly common as MR imaging applications grow. Other newer techniques abandon the Cartesian (row and column) approach to covering k-space and substitute spiral or radial trajectories, each providing both advantages and disadvantages that have relevance to particular applications. Still newer techniques known collectively as parallel imaging[12,13] exploit the signals from an array of coils and combine them with sophisticated mathematics to effectively conduct, essentially, a number of imaging experiments at the same time. The increase in available signal can be used to improve quality or can be traded off for speed.

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