

MAGNETIC RESONANCE IMAGING (MRI)

The importance of magnetic resonance imaging (MRI) in clinical imaging has exceeded even the most optimistic hopes of researchers from the 1980s. An ability to manipulate and adjust tissue contrast with increasingly complex pulse sequences is at the crux of this success, as described in the previous chapter. However, the ability to accurately determine the position from the nuclear magnetic resonance (NMR) signal and thus create an image is the basis of clinical MRI. This chapter describes how the MR signals are localized in three dimensions, and the relationship between frequency and spatial position is examined. Characteristics of the MR image, including factors that affect the signal-to-noise ratio (SNR), blood flow effects, and artifacts are described. Concluding the chapter is an overview of MR equipment, quality control procedures, and biologic safety issues with respect to magnetic fields and radiofrequency (RF) energy.

15.1 LOCALIZATION OF THE MR SIGNAL

The protons in a material, with the use of an external uniform magnetic field and RF energy of specific frequency, are excited and subsequently produce signals with amplitudes dependent on relaxation characteristics and spin density, as previously discussed (see Chapter 14). Spatial localization, fundamental to MR imaging, requires the imposition of magnetic field nonuniformities—magnetic gradients superimposed upon the homogeneous and much stronger main magnetic field, which are used to distinguish the positions of the signal in a three-dimensional object (the patient). Conventional MRI involves RF excitations combined with magnetic field gradients to localize the signal from individual volume elements (voxels) in the patient.

Magnetic Field Gradients

Magnetic fields with predictable directionality and strength are produced in a coil wire energized with a direct electric current of specific polarity and amplitude. Magnetic field gradients are obtained by superimposing the magnetic fields of one or more coils with a precisely defined geometry (Fig. 15-1). With appropriate design, the gradient coils create a magnetic field that linearly varies in strength versus distance over a pre-

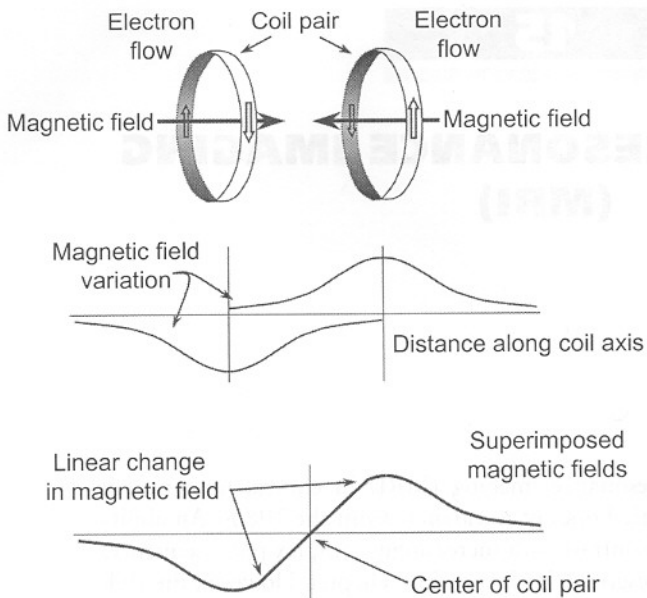


FIGURE 15-1. Individual conducting wire coils that are separately energized with currents of opposite directions produce magnetic fields of opposite polarity. Magnetic field strength reduces with distance from the center of each coil. When combined, the magnetic field variations form a linear change between the coils, producing a linear magnetic field gradient.

defined field of view (FOV). When superimposed upon a homogeneous magnetic field (e.g., the main magnet, B_0), positive gradient field adds to B_0 and negative gradient field reduces B_0 . Inside the magnet bore, three sets of gradients reside along the coordinate axes— x , y , and z —and produce a magnetic field variation determined by the magnitude of the applied current in each coil set (Fig. 15-2). When independently energized, the three coils (x , y , z) produce a linearly variable magnetic field in any direction, where the net gradient is equal to $\sqrt{G_x^2 + G_y^2 + G_z^2}$. Gradient polarity reversals (positive to negative and negative to positive changes in magnetic field strength) are achieved by reversing the current direction in the gradient coils.

Two properties of gradient systems are important: (a) The peak amplitude of the gradient field determines the “steepness” of the gradient field. Gradient magnetic field strength typically ranges from 1 to 50 millitesla per meter (mT/m) [0.1 to 5 gauss (G)/cm]. (b) The slew rate is the time required to achieve the peak magnetic field amplitude, where shorter time is better. Typical slew rates of gradient fields are from 5 mT/m/msec to 250 mT/m/msec. Limitations in the slew rate are caused by eddy currents, which are electric currents induced in nearby conductors that oppose the generation of the gradient field (e.g., both the RF coils and the patient produce eddy currents). Actively shielded gradient coils and compensation circuits reduce the problems introduced by eddy currents.

The gradient is a linear, position-dependent magnetic field applied across the FOV, and it causes protons to alter their precessional frequency corresponding to their position along the applied gradient in a known and predictable way. At the middle of the gradient exists the null, where no change in the net magnetic field or precessional frequency exists. A linear increase (or decrease) in precessional frequency occurs with the variation of the local magnetic field strength away from the null (Fig. 15-3 and Table 15-1). Location of protons along the gradient is determined by their frequency and phase; the gradient amplitude and the number of samples over the FOV determine the frequency bandwidth (BW) across each pixel.

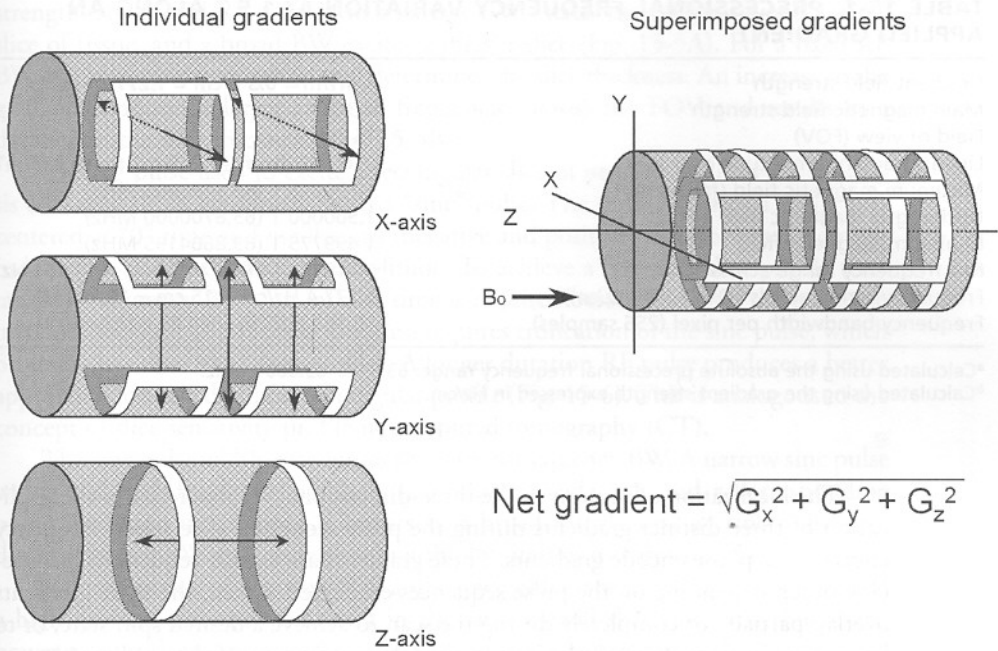


FIGURE 15-2. Within a large stationary magnetic field, field gradients are produced by three separate saddle coils placed within the central core of the magnet, along the x, y, or z direction. Magnetic field gradients of arbitrary direction are produced by the vector addition of the individual gradients turned on simultaneously. Any gradient direction is possible by superimposition of the three-axis gradient system.

The Larmor equation ($\omega = \gamma B$) allows the gradient amplitude (for protons) to be expressed in the units of Hz/cm. For instance, a 10-mT/m gradient can be expressed as $10 \text{ mT/m} \times 42.58 \text{ MHz/T} \times 1\text{T}/1,000 \text{ mT} = 0.4258 \text{ MHz/m}$, which is equivalent to 425.8 kHz/m or 4258 Hz/cm. This straightforward description of the gradient strength facilitates determining the frequency BW across the FOV, independent of the main magnetic field strength.

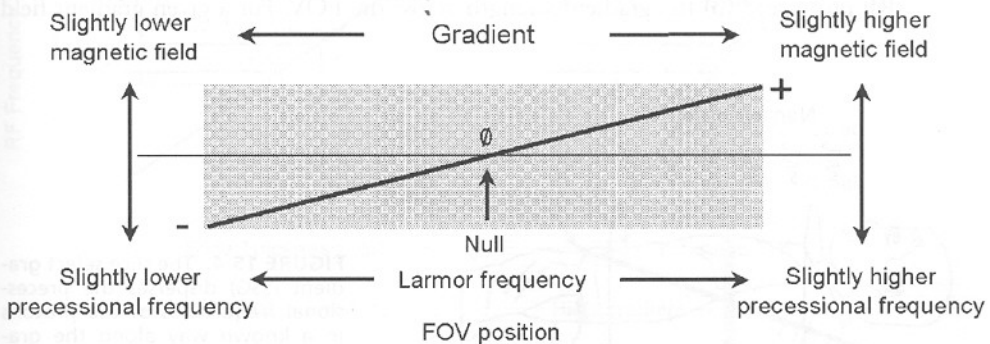


FIGURE 15-3. The gradient field creates a net positive and negative magnetic environment that adds to and subtracts from the main magnetic field. Associated with the local change in magnetic field is a local change in precessional frequencies, per the Larmor equation. The frequencies thus directly vary across the field in proportion to the applied gradient strength.

TABLE 15-1. PRECESSIONAL FREQUENCY VARIATION AT 1.5 T ALONG AN APPLIED GRADIENT

Gradient field strength	3 mT/m = 0.3 g/cm = 1,277.4 Hz/cm
Main magnetic field strength	1.5 T
Field of view (FOV)	15 cm
Linear gradient amplitude over FOV	0.45 mT; from -0.225 mT to +0.225 mT
Maximum magnetic field (frequency)	1.500225 T (63.8795805 MHz)
Unchanged magnetic field at null	1.500000 T (63.8700000 MHz)
Minimum magnetic field	1.499775 T (63.8604195 MHz)
Net frequency range across FOV ^a	0.019161 MHz = 19.2 kHz = 19,161 Hz
Frequency range across FOV (1,278 Hz/cm) ^b	1,277.4 Hz/cm $\times$ 15 cm = 19,161 Hz
Frequency bandwidth per pixel (256 samples)	19,161 Hz/256 = 74.85 Hz/pixel

^aCalculated using the absolute precessional frequency range: 63.8796–63.8604 MHz.

^bCalculated using the gradient strength expressed in Hz/cm.

The localization of protons in the three-dimensional volume requires the application of three distinct gradients during the pulse sequence: slice select, frequency encode, and phase encode gradients. These gradients are usually sequenced in a specific order, depending on the pulse sequences employed. Often, the three gradients overlap partially or completely during the scan to achieve a desired spin state, or to leave spins in their original phase state after the application of the gradient(s).

Slice Select Gradient

The RF antennas that produce the RF pulses do not have the ability to spatially direct the RF energy. Thus, the slice select gradient (SSG) determines the slice of tissue to be imaged in the body, in conjunction with the RF excitation pulse. For axial MR images, this gradient is applied along the long (cranial-caudal) axis of the body. Proton precessional frequencies vary according to their distance from the null of the SSG. A selective frequency (narrow band) RF pulse is applied to the whole volume, but only those spins along the gradient that have a precessional frequency equal to the frequency of the RF will absorb energy due to the resonance phenomenon (Fig. 15-4).

Slice thickness is determined by two parameters: (a) the bandwidth (BW) of the RF pulse, and (b) the gradient strength across the FOV. For a given gradient field

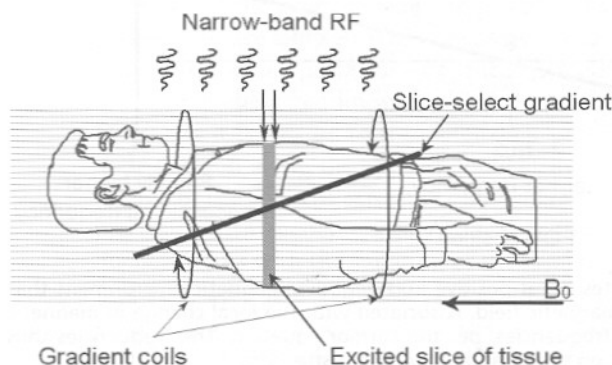


FIGURE 15-4. The slice select gradient (SSG) disperses the precessional frequencies of the protons in a known way along the gradient. A narrow-band radiofrequency (RF) pulse excites only a selected volume (slice) of tissues, determined by RF bandwidth and SSG strength.

strength, an applied RF pulse with a narrow BW excites the protons over a narrow slice of tissue, and a broad BW excites a thicker slice (Fig. 15-5A). For a fixed RF BW, the SEG field strength (slope) determines the slice thickness. An increase in the gradient produces a larger range of frequencies across the FOV and results in a decrease in the slice thickness (Fig. 15-5B).

The RF pulse used to excite a rectangular slice of protons requires the synthesis of a specialized waveform called a “sinc” pulse. The pulse contains a main lobe centered at “0” time and oscillations (negative and positive) that decrease in amplitude before and after the peak amplitude. To achieve a “perfect” slice profile (rectangular frequency BW), an infinite time is required before and after the pulse, an unachievable goal. Short pulse duration requires truncation of the sinc pulse, which produces less than ideal slice profiles. A longer duration RF pulse produces a better approximation to a desirable rectangular profile (Fig. 15-6). This is analogous to the concept of slice sensitivity profile in computed tomography (CT).

The sinc pulse width determines the output frequency BW. A narrow sinc pulse width and high-frequency oscillations produce a wide BW and a corresponding broad excitation distribution. Conversely, a broad, slowly varying sinc pulse produces a narrow BW and corresponding thin excitation distribution (Fig. 15-7).

A combination of a narrow BW and a low gradient strength or a wide BW and a high gradient strength can result in the same overall slice thickness. There are, however, considerations regarding the SNR of the *acquired* data as

$$\text{SNR} \propto \frac{1}{\sqrt{\text{BW}}}.$$

A narrow BW results in increased SNR. By decreasing the BW by a factor of four, the SNR is increased by a factor of two, as long as the same slice thickness is acquired by decreasing the gradient strength. Narrow BW is not always the appropriate choice, however, because chemical shift artifacts and other undesirable image

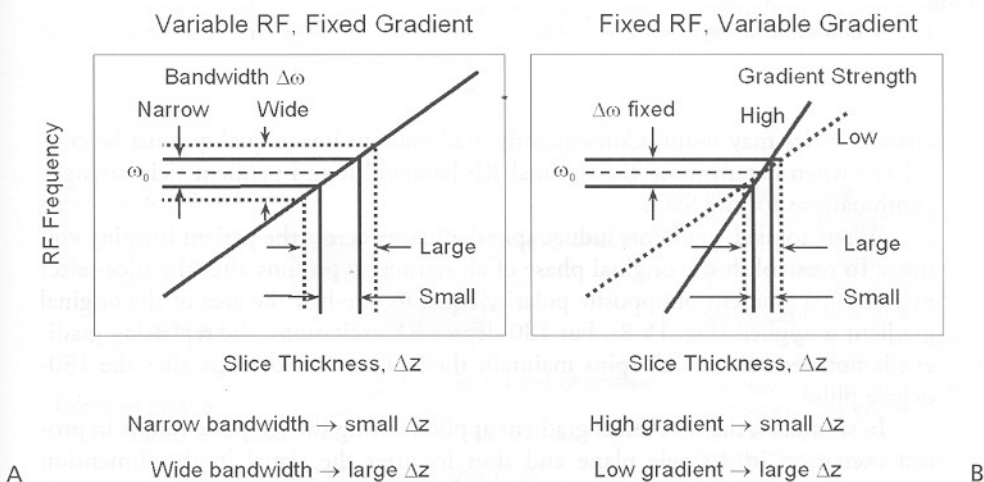


FIGURE 15-5. Slice thickness is dependent on RF bandwidth and gradient strength. **A:** For a fixed gradient strength, the RF bandwidth determines the slice thickness. **B:** For a fixed RF bandwidth, gradient strength determines the slice thickness.

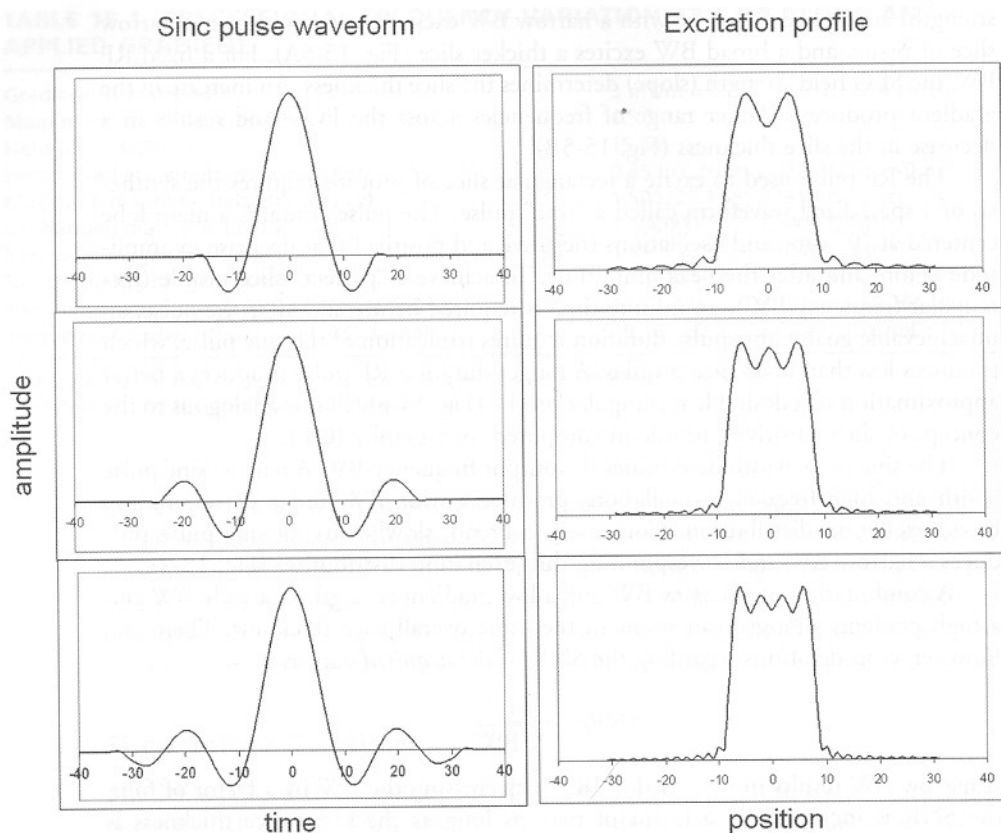


FIGURE 15-6. An approximate rectangular slice profile of precessing protons is achieved with the use of a RF waveform known as a "sinc" pulse, which evolves with oscillatory tails of increasing amplitude to the main lobe at time = 0, and decays in a symmetric manner with time. Truncation to limit the RF pulse duration produces nonrectangular slice profiles. As the RF pulse duration increases, the slice profile (inclusion of more lobes) is increased. The graphs have relative units for time, distance, and amplitude.

characteristics may result. Consequently, trade-offs in image quality must be considered when determining the optimal RF bandwidth and gradient field strength combinations for the SSG.

When applied, gradients induce spin dephasing across the patient imaging volume. To reestablish the original phase of all stationary protons after the slice-select excitation, a gradient of opposite polarity equal to one-half the area of the original gradient is applied (Fig. 15-8). For 180-degree RF excitations, the rephasing gradient is not necessary, as all spins maintain their phase relationships after the 180-degree pulse.

In summary, the slice select gradient applied during the RF pulse results in proton excitation in a single plane and thus localizes the signal in the dimension orthogonal to the gradient. It is the first of three gradients applied to the sample volume.

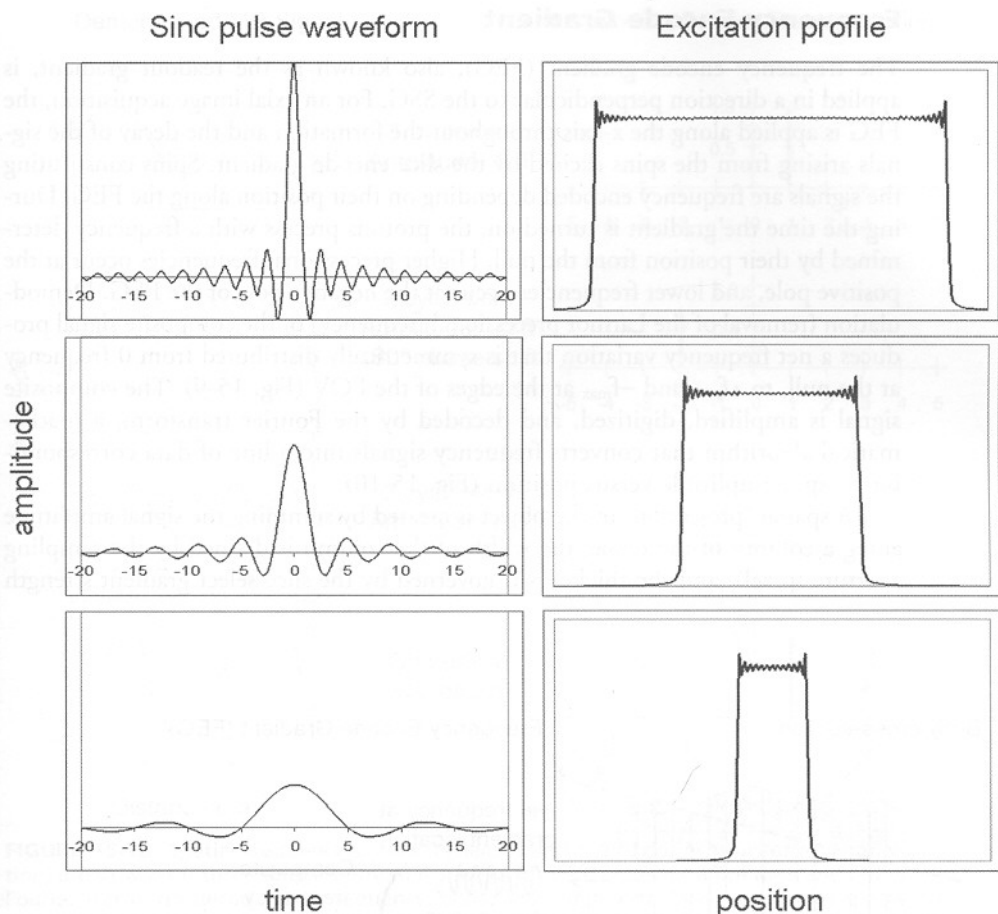


FIGURE 15-7. The width of the sinc pulse lobes determines the RF bandwidth. Narrow lobes produce wide bandwidth; wide lobes produce narrow bandwidth. In these examples, the area under the main sinc pulse lobe of each is approximately equal, resulting in equal amplitude of the corresponding spatial excitation characteristics. The graphs have relative units for time, distance and amplitude.

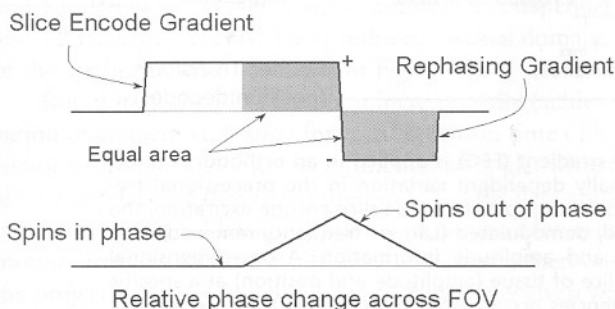


FIGURE 15-8. Gradients produce a range of precessional frequencies within the volume of tissue experiencing the magnetic field variations, which leaves the spins dephased after the gradient is removed. A rephasing gradient of opposite polarity serves to rewind the spins back to a coherent phase relationship when its area is equal to one-half of the total area of the original gradient.

Frequency Encode Gradient

The frequency encode gradient (FEG), also known as the readout gradient, is applied in a direction perpendicular to the SSG. For an axial image acquisition, the FEG is applied along the x-axis throughout the formation and the decay of the signals arising from the spins excited by the slice encode gradient. Spins constituting the signals are frequency encoded depending on their position along the FEG. During the time the gradient is turned on, the protons precess with a frequency determined by their position from the null. Higher precessional frequencies occur at the positive pole, and lower frequencies occur at the negative pole of the FEG. Demodulation (removal of the Larmor precessional frequency) of the composite signal produces a net frequency variation that is symmetrically distributed from 0 frequency at the null, to $+f_{\max}$ and $-f_{\max}$ at the edges of the FOV (Fig. 15-9). The composite signal is amplified, digitized, and decoded by the Fourier transform, a mathematical algorithm that converts frequency signals into a line of data corresponding to spin amplitude versus position (Fig. 15-10).

A spatial "projection" of the object is created by summing the signal amplitude along a column of the tissue; the width of the column is defined by the sampling aperture (pixel), and the thickness is governed by the slice select gradient strength

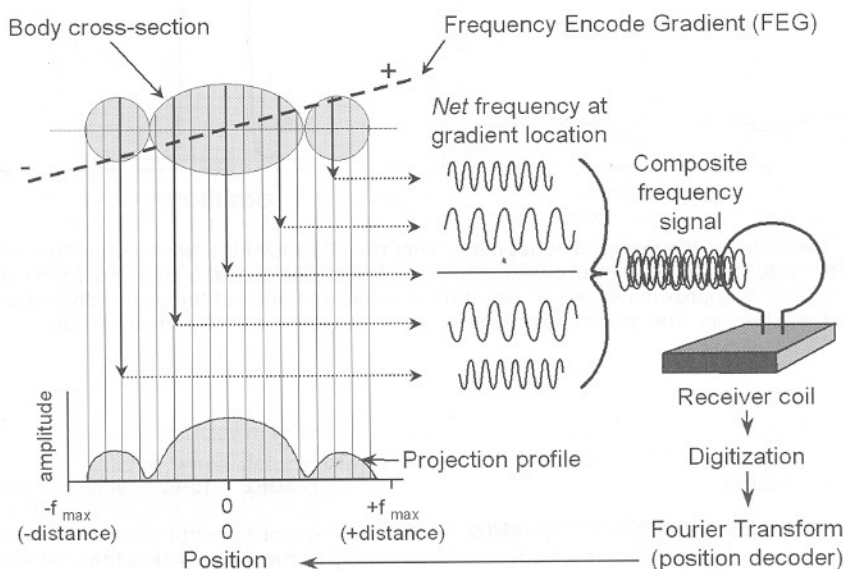


FIGURE 15-9. The frequency encode gradient (FEG) is applied in an orthogonal direction to the SSG, and confers a spatially dependent variation in the precessional frequencies of the protons. Acting only on those spins from the slice encode excitation, the composite signal is acquired, digitized, demodulated (Larmor frequency removed), and Fourier transformed into frequency and amplitude information. A one-dimensional array represents a projection of the slice of tissue (amplitude and position) at a specific angle. (Demodulation into net frequencies occurs after detection by the receiver coil; this is shown in the figure for clarity only.)

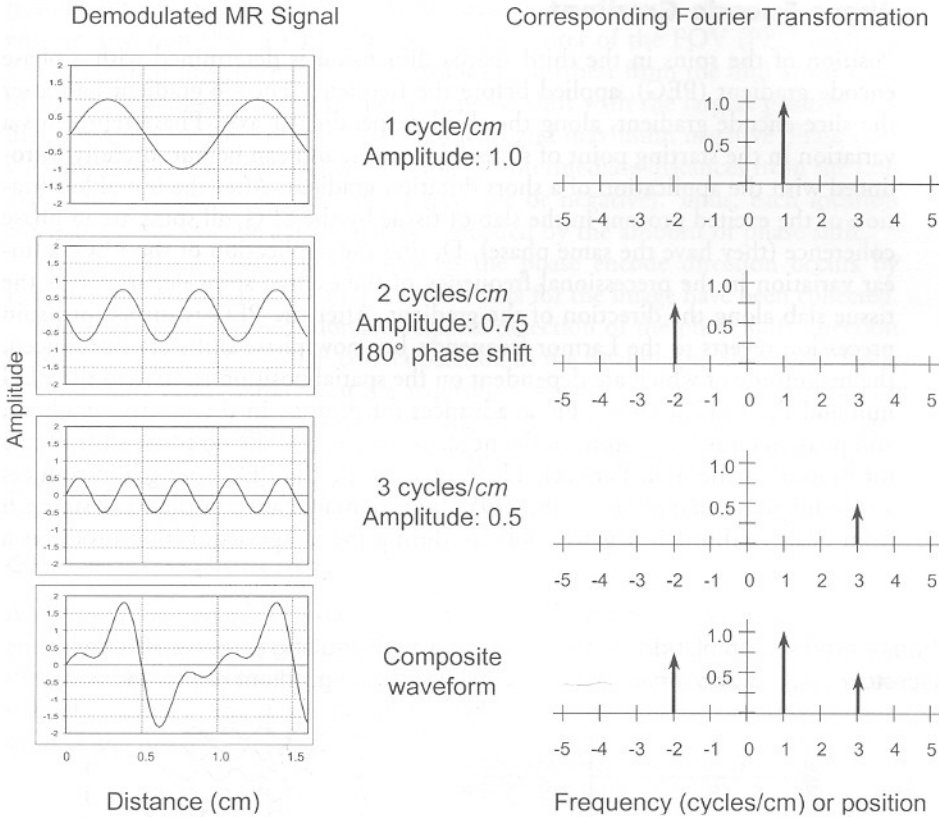


FIGURE 15-10. Spatial frequency signals (cycle/cm) and their Fourier transforms (spatial position) are shown for three simple sinusoidal waveforms with a specific amplitude and phase. The Fourier transform decodes the frequency, phase and amplitude variations in the spatial frequency domain into a corresponding position and amplitude in the spatial domain. A 180-degree phase shift (**2nd from top**) is shifted in the negative direction from the origin. The composite waveform (a summation of all waveforms, **lower left**) is decoded by Fourier transformation into the corresponding positions and amplitudes (**lower right**).

and RF bandwidth. Overall signal amplitude is dependent on spin density and T1 and T2 relaxation events. This produces a spatial domain profile along the direction of the applied gradient (bottom of Fig. 15-9).

Rotation of the FEG direction incrementally (achieved with simultaneous activation of gradient coil pairs) for each repetition time (TR) interval can provide projections through the object as a function of angle, which is very similar conceptually to the data set that is acquired in CT. This data could produce a tomographic image using filtered backprojection techniques. However, because of sensitivity to motion artifacts and the availability of alternate methods to generate MR images, the projection-reconstruction method has largely given way to phase encoding techniques.

Phase Encode Gradient

Position of the spins in the third spatial dimension is determined with a phase encode gradient (PEG), applied before the frequency encode gradient and after the slice encode gradient, along the third perpendicular axis. Phase represents a variation in the starting point of sinusoidal waves, and can be purposefully introduced with the application of a short duration gradient. After the initial localization of the excited protons in the slab of tissue by the SEG, all spins are in phase coherence (they have the same phase). During the application of the PEG, a linear variation in the precessional frequency of the excited spins occurs across the tissue slab along the direction of the gradient. After the PEG is turned off, spin precession reverts to the Larmor frequency, but now phase shifts are introduced, the magnitude of which are dependent on the spatial position relative to the PEG null and the PEG strength. Phase advances for protons in the positive gradient, and phase retards for protons in the negative gradient, while no phase shift occurs for protons at the null. For each TR interval, a specific PEG strength introduces a specific phase change across the FOV. Incremental change in the PEG strength from positive through negative polarity during the image acquisition produces a

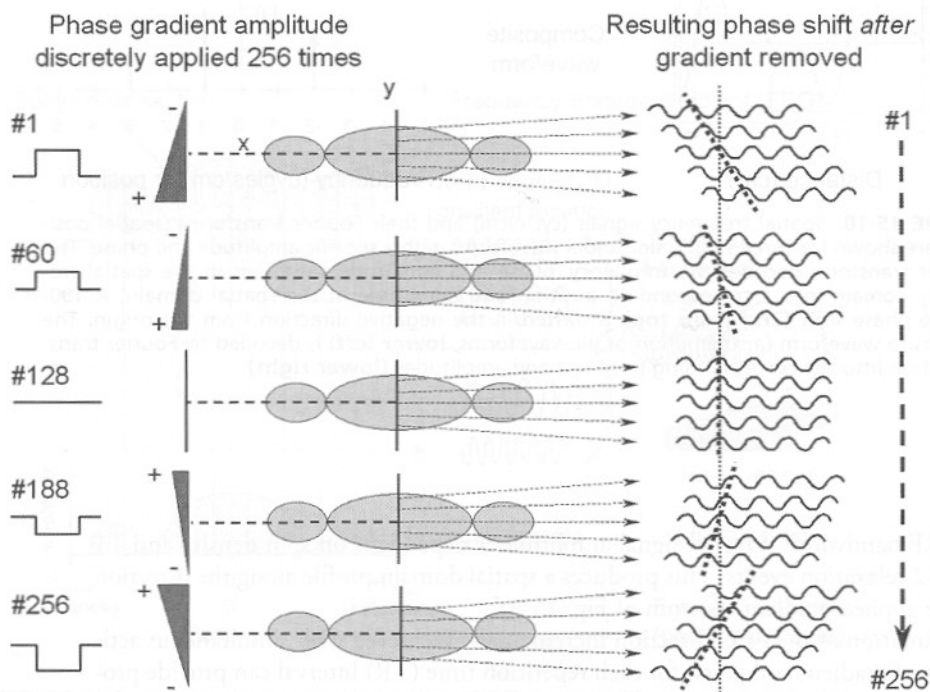


FIGURE 15-11. The phase encode gradient (PEG) produces a spatially dependent variation in angular frequency of the excited spins for a brief duration, and generates a spatially dependent variation in phase when the spins return to the Larmor frequency. Incremental changes in the PEG strength for each TR interval spatially encodes the phase variations: protons at the null of the PEG do not experience any phase change, while protons in the periphery experience a large phase change dependent on the distance from the null. The incremental variation of the PEG strength can be thought of as providing specific "views" of the three-dimensional volume because the SSG and FEG remain fixed throughout the acquisition.

positionally dependent phase shift at each position along the applied phase encode direction (Fig. 15-11). Protons at the center of the FOV (PEG null) do not experience any phase shift. Protons located furthest from the null at the edge of the FOV gain the maximum positive phase shift with the largest positive gradient, no phase shift with the “null” gradient, and maximum negative phase shift with the largest negative gradient. Protons at intermediate distances from the null experience intermediate phase shifts (positive or negative). Thus, each location along the phase encode axis is spatially encoded by the amount of phase shift.

Decoding the spatial position along the phase encode direction occurs by Fourier transformation, only after all of the data for the image have been collected. Symmetry in the frequency domain requires detection of the phase shift direction (positive or negative) to assign correct position in the final image. Since the PEG is incrementally varied throughout the acquisition sequence (e.g., 128 times in a 128×128 MR image), slight changes in position caused by motion will cause a corresponding change in phase, and will be manifested as partial (artifactual) copies of anatomy displaced along the phase encode axis.

Gradient Sequencing

A spin echo sequence is illustrated in Fig. 15-12, showing the timing of the gradients in conjunction with the RF excitation pulses and the data acquisition during the evolution and decay of the echo. This sequence is repeated with slight incremental changes in the phase encode gradient strength to define the three-dimensions in the image.

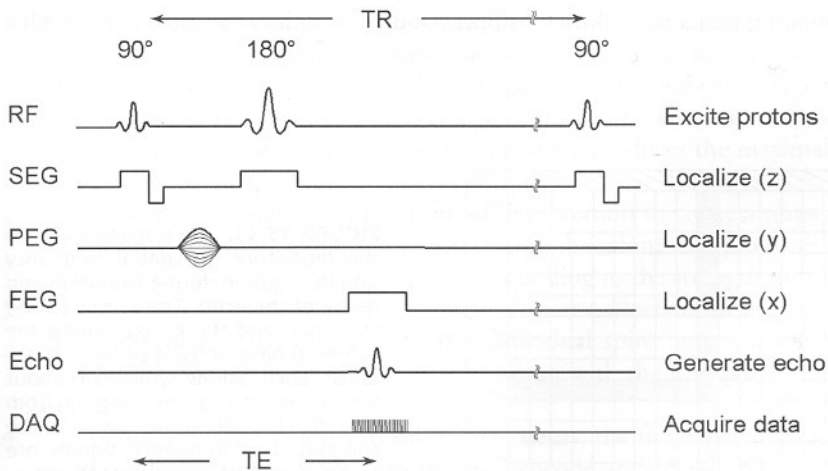


FIGURE 15-12. A typical spin-echo pulse sequence diagram indicates the timing of the SSG, PEG, and FEG during the repetition time (TR) interval, synchronized with the RF pulses and the data acquisition (DAQ) when the echo appears. Each TR interval is repeated with a different PEG strength (this appears as multiple lines in the illustration, but only one PEG strength is applied as indicated by the *bold line* in this example).