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### Index terms:

Magnetic resonance (MR), echo planar, \*\*.121412<sup>2</sup>  
Magnetic resonance (MR), fat suppression, \*\*.121415  
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### Abbreviations:

FLAIR = fluid-attenuated inversion recovery  
PRESS = point-resolved spectroscopy  
RF = radio frequency  
STEAM = stimulated echo acquisition mode  
STIR = short inversion time inversion recovery  
TE = echo time  
TI = inversion time  
TR = repetition time  
2D = two-dimensional  
3D = three-dimensional

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# MR Imaging Abbreviations, Definitions, and Descriptions: A Review<sup>1</sup>

Magnetic resonance (MR) imaging is a powerful imaging modality that combines excellent soft-tissue contrast and spatial resolution. One of the strengths of MR imaging is the myriad of measurement techniques, known as pulse sequences and parameter modifications, available for use. The variety of MR imaging methods has also introduced complexity, as there exists no standard for technique nomenclature that can be used by manufacturers. Early in the development of MR, basic MR imaging terminology was standardized by the American College of Radiology (1). However, many imaging methods and variations have since been developed. As manufacturers implement new techniques, they develop vendor-specific nomenclature for them. This has made it extremely difficult for the radiologist or technologist to compare results acquired with different MR systems or to ensure that the proper images are obtained.

This review is intended to provide the reader with a brief description of the general categories of MR pulse sequences and measurement techniques as implemented on MR imaging systems from some of the major manufacturers. Our assumption in writing this article is that the reader is familiar with the basic concepts of MR imaging as used on commercially available MR systems. There are many introductory books currently available that describe the fundamental principles of MR imaging in more detail (2–6). We used the applications guides as supplied by the individual manufacturers as the primary references for vendor-specific implementation of the pulse sequences (7–14). This provided a nonproprietary, public, and accurate source of the particular techniques as implemented by the manufacturers. The applications guides that were used provided the most detailed information on pulse sequence definitions, even though they were not necessarily the most recent guides. Our goal is not to critique the particular implementation of a sequence or technique by any specific manufacturer. Rather, this review is an attempt to provide both a general description of some common pulse sequences, modifications, and terminology and a reference to the practicing radiologist or technologist when comparing studies from different MR systems.

## PULSE SEQUENCES AND TIMING DIAGRAMS

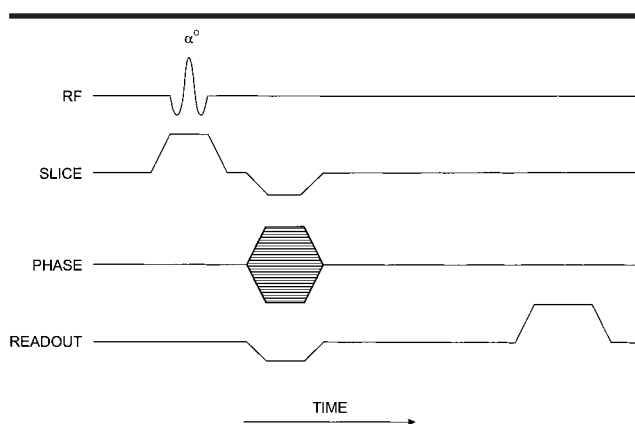
The heart of an MR measurement is the technique used for data collection, known as a pulse sequence. Pulse sequences are computer programs that control all hardware aspects of the measurement process. Comparisons of hardware activity between different pulse sequences can be easily made through the use of timing diagrams. These are schematic figures often used to illustrate the basic hardware steps that are incorporated into a pulse sequence. Although there may be stylistic differences between diagrams by different authors, the general features are the same for all diagrams. Time during sequence execution is indicated along the horizontal axis. Each line corresponds to a different hardware component. The vertical separation between each line is employed only for visualization. At a minimum, four lines are needed to completely describe any pulse sequence: one for the radio-frequency (RF) transmitter and one for each gradient (indicated as  $G_x$ ,  $G_y$ ,  $G_z$ , or  $G_{\text{section}}$ ,  $G_{\text{phase}}$ ,  $G_{\text{readout}}$ ). Additional lines may be added to indicate other activity such as analog-to-digital converter, or ADC, sampling. Activity for a particular component such as a gradient pulse is shown as a deviation above or below the horizontal line. Simultaneous activity from more than one component such as the RF pulse and section-selection gradient is indicated as nonzero activity from both lines at the same horizontal position. Constant amplitude gradient pulses are shown as simple deviations from zero. Gradient amplitudes that change during the measurement such as for phase encoding are represented as hatched regions.

The hardware steps of a pulse sequence are performed in repeated blocks of instructions known as loops. The total time illustrated by a timing diagram is generally the smallest such block, known as the section loop or sequence loop. This is the minimum repeat time required to complete one loop through all hardware steps per the given repeat unit (section, phase-encoding step, acquisition, etc). The order in which the steps are incremented is sequence dependent, but there are three loop structures that are typically used: two-dimensional (2D) multisection, 2D sequential section, and three-dimensional (3D). Two-dimensional multisection (or multislice) is the most common loop structure. Narrow regions of tissue (~2–10 mm) are excited with each excitation pulse, and the measured signal for each image is produced from only this volume of tissue. Each section is excited once per repetition time (TR) before signal averaging or changing the phase-encoding gradient amplitude (Fig 1).

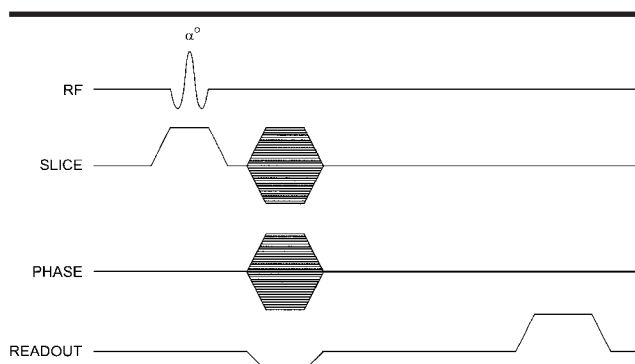
The 2D multisection loop structure provides the most time-efficient method for acquiring information from individual sections in that data from each section are acquired every TR. The measurement time (also referred to as "scan time"),  $T$ , is calculated as follows:  $T_{\text{multisection}} = \text{TR} \times (\text{no. of acquisitions}) \times (\text{no. of phase-encoding steps})$ .

The 2D sequential section looping uses narrow volume excitation and detection like that of the 2D multisection loop but acquires all information from a single section (eg, phase encoding, acquisitions) before advancing to the next section position. This is used for special applications and with very short TR. The measurement time for sequential section looping is  $T_{\text{sequential}} = \text{TR} \times (\text{no. of acquisitions}) \times (\text{no. of phase-encoding steps}) \times (\text{no. of sections})$ .

The third loop structure is 3D or volume imaging. For 3D imaging, large volumes of tissue (typically 30–200 mm or larger) are excited by each excitation pulse. The excitation volume is subsequently phase encoded in directions perpendicular (phase encoding) and parallel (partitions or section encoding) to the plane of excitation. In contrast to 2D imaging in which there is only a single gradient table in the phase-encoding direction, 3D imaging uses two independent gradient tables in the section and phase-encoding directions (Fig 2). The section encoding of the volume enables thin, contiguous sections with excellent signal-to-noise ratio to be produced on the basis of the signal from the excited



**Figure 1.** A 2D pulse sequence timing diagram. Fixed amplitude pulses are indicated as simple deviations from horizontal. Variable amplitude pulses are indicated as hatched regions. Section (*slice*) selection and signal detection are repeated in amplitude, duration, and relative timing each time the sequence is executed. A single phase-encoding table is present, which is incremented in amplitude each time the sequence is executed.



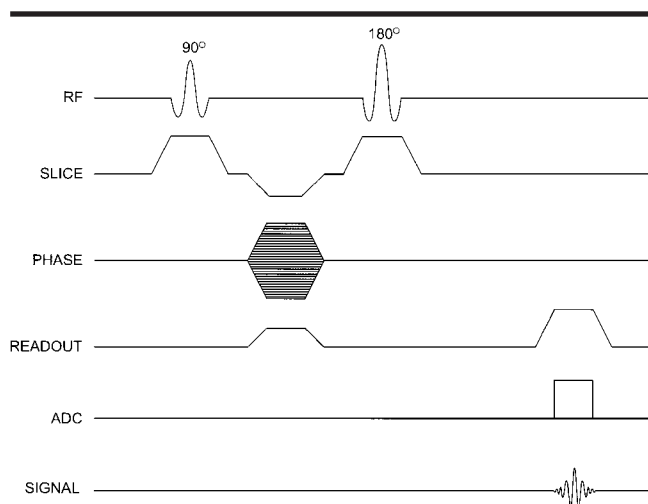
**Figure 2.** A 3D pulse sequence timing diagram. Volume excitation and signal detection are repeated in amplitude, duration, and relative timing each time. Two phase-encoding tables are present, one in the phase-encoding direction and one in the section (*slice*) direction, which are independently incremented in amplitude each time the sequence is executed. The compensation gradient in the section direction is incorporated into the gradient table.

volume. The measurement time for volume excitation techniques is  $T_{3D} = \text{TR} \times (\text{no. of acquisitions}) \times (\text{no. of phase-encoding steps}) \times (\text{no. of 3D partitions})$ .

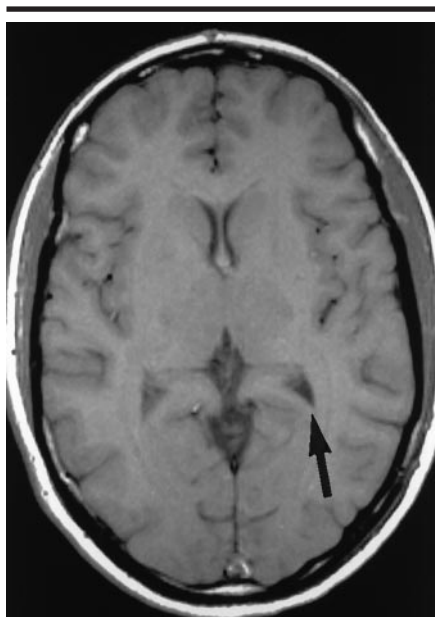
Techniques for reducing the measurement time for a particular measurement technique involve altering one or more of the measurement parameters that appear in the appropriate equation. There are practical limits to the amount of measurement time reduction possible, since the image contrast-to-noise (signal difference-to-noise) ratio and spatial resolution as well as the measurement time will be affected by the various parameters. For example, reduction of TR increases the amount of T1 saturation contributing to the contrast, as well as decreases the number of sections that can be obtained

by using a multisection loop. Reducing the number of acquisitions does not affect the intrinsic signal intensity or contrast on the image, but it does increase the relative amount of noise in each line of data. In addition, for standard imaging techniques, there is a minimum of one acquisition required for each image. The most common parameter that is adjusted to reduce the measurement time is the number of phase-encoding steps. In a similar fashion, the number of partitions for a 3D acquisition may be adjusted.

Reducing the number of phase-encoding steps for the measurement can be done in two ways, both of which maintain the field of view for the image. One approach increases the size of each volume element (voxel) within the image.



**Figure 3.** Single-echo spin-echo pulse sequence timing diagram. A pair of RF pulses produces a single echo. This echo is always detected in the presence of a readout gradient of constant amplitude. The excitation-detection process is repeated many times, each time with a different amplitude of the phase-encoding gradient applied prior to signal detection. ADC = analog-to-digital conversion.



**Figure 4.** Transverse T1-weighted spin-echo image (500/15 [TR msec/TE msec]) shows that the cerebrospinal fluid in the lateral ventricles has low signal intensity (arrow).

This is accomplished by acquiring a narrower range of phase-encoding steps while keeping constant the change from one step to the next, acquiring the raw data set in a coarser fashion. The “missing” data are replaced with zeroes prior to reconstruction. The signal-to-noise ratio is increased in each voxel since more tissue is included, but the spatial resolution is decreased. The particular clinical

application will dictate the amount of resolution loss that is acceptable. Alternatively, the number of phase-encoding steps can be reduced while maintaining the spatial resolution by keeping the maximum phase-encoding gradient amplitude equal to that of the full matrix. Reducing the number of phase-encoding steps reduces the number of independent measurements, allowing the noise to be a greater percentage of the total result. The raw data matrix will be incomplete and asymmetric, necessitating corrections to be made prior to image reconstruction. These corrections generally make the images more sensitive to motion than is the corresponding full acquisition. This is the basis for the class of techniques known as partial Fourier or partial k-space acquisition approaches (15).

### CLASSES OF PULSE SEQUENCES

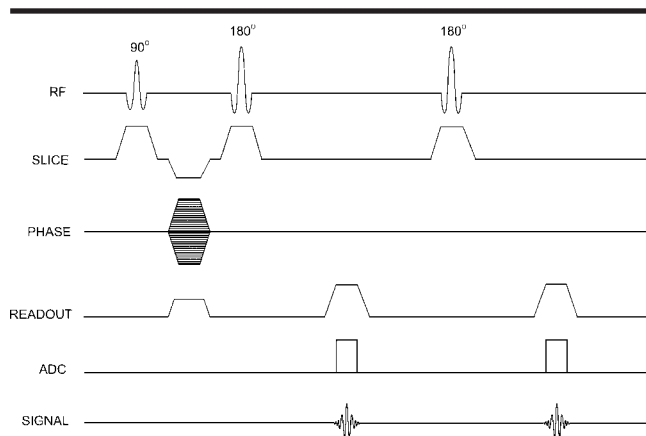
Two of the most confusing aspects of MR imaging are the large number of pulse sequences that are possible and the different manners in which different vendors have implemented and named the same technique. In spite of this, there are similarities between pulse sequences that make a general categorization possible. First, some form of echo signal is detected. Each measurement will have an operator-selectable parameter, the echo time (TE), defined as the time from the excitation pulse to the echo maximum. The second common feature of pulse

sequences is that the basic aspects of spatial localization are used in all imaging sequences currently used in MR imaging. Some form of region-selective excitation is used. Localization of the RF energy to a volume of tissue is accomplished by using frequency selective, narrow bandwidth pulses in conjunction with a gradient called the section-selection gradient. All signals are detected in the presence of a gradient known as the readout or frequency-encoding gradient, which is oriented perpendicular to the section-selection gradient. There will be some form of residual phase advancement or retardation based on the position of the tissue in the third direction, the phase-encoding direction, which is perpendicular to the other two directions. Finally, gradient pulses are usually applied in complementary pairs. This allows any phase variation induced by the gradient to be reversed. For most MR pulse sequences, these complementary pulse pairs are applied in the section-selection and readout directions. The following sections present the major classes of pulse sequences that are currently used in MR imaging and their defining characteristics.

At the end of the text are tables that describe common measurement techniques and acronyms, and the manufacturers that use them. Certain generic terms are used by all manufacturers such as magnetization transfer, or MT. Other terms are used by specific manufacturers to identify certain aspects or features of sequences. In addition, depending on the software of the manufacturer, certain techniques may be identified as separate techniques or as options under more general techniques. Also, no indication is made regarding 2D and 3D acquisition, nor between sequential versus multisection mode. Specific questions about particular sequences should be addressed to the individual manufacturer. Although each table is cited within the following sections, they are grouped together at the end of the text to provide a stand-alone reference document for the reader.

### SPIN ECHO

The most common class of pulse sequences is based on detection of a spin or Hahn echo. Spin-echo pulse sequences are characterized by the use of an excitation pulse (often called the alpha [ $\alpha$ ] pulse) and one or more 180° pulses following excitation to refocus the transverse magnetization (16). The refocusing pulses generally excite the same volume of tis-



**Figure 5.** Multiple-echo spin-echo pulse sequence timing diagram, two echoes illustrated. Multiple refocusing RF pulses are used to produce multiple echoes. Each echo is detected in the presence of a constant amplitude readout gradient, following a common amplitude of the variable phase-encoding gradient. ADC = analog-to-digital conversion.

sue as the excitation pulse. Each refocusing pulse produces an echo known as a spin echo. The differences between the types of spin-echo sequences are in the number of refocusing pulses and the number of phase-encoding tables used in the measurement. Table 1 provides a list of the common types of spin-echo pulse sequences that are currently available.

### Single-Echo Spin Echo

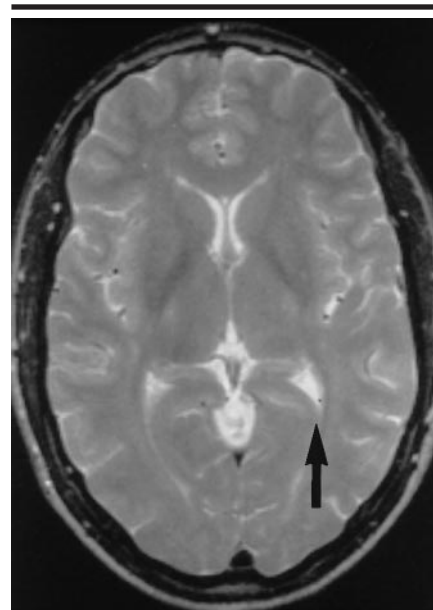
Single-echo spin-echo sequences use a section excitation pulse with one 180° refocusing pulse per section loop, using a multisection or sequential section loop structure. A single phase-encoding gradient amplitude is applied per RF excitation pulse (Fig 3). Each echo is measured from the excited section at the selected TE following application of the phase-encoding gradient pulse. After reconstruction, amplitude variations in the image result from differences in tissue-specific properties (T1, T2, proton density, or flow velocity). Single-echo spin-echo sequences are most often used to produce T1-weighted images. This is accomplished by using relatively short TRs and TEs for data collection. Typical parameters for T1-weighted single-echo spin-echo imaging are a TR of 500 msec and a TE of 15 msec. Observation that fluid-containing structures have very low signal intensity is generally an indication that the particular sequence employed is T1-weighted (Fig 4).

### Multiecho Spin Echo

Standard multiecho sequences have the same basic structure as single-echo se-

quences except that there is more than one 180° refocusing pulse per section loop (Fig 5). Each refocusing pulse produces a spin echo, each one occurring at a different TE. A single phase-encoding gradient is applied per excitation pulse, so that each echo differs in the amount of T2 weighting present. All phase-encoding steps used to reconstruct an image are acquired at the same TE. Multiecho sequences are most commonly used to produce intermediate-weighted (short TE) and T2-weighted (long TE) images when TR is sufficient to allow for relatively complete T1 relaxation. Typical parameters for multiecho spin-echo images are TR of 2,000–4,500 msec and TE of 20 msec (intermediate-weighted) and 90–140 msec (T2-weighted). Observation that fluid-containing structures have high signal intensity is generally an indication that the sequence is T2-weighted (Fig 6).

There are a few common implementations of multiecho sequences. Some variations have the later TEs as multiples of the earlier TEs, though this is not a requirement. One advantage for having the second TE as a multiple of the first TE is the compensation of pulse imperfections by the second 180° pulse. This is a phenomenon known as even echo rephasing and can result in improved signal intensity on the second-echo image. Improved RF pulse design can reduce the pulse imperfections to a minimum so that the improvement in signal intensity is marginal. Another variation is in the total number of images that are acquired. While two echoes are most commonly acquired in multiecho sequences, more than two echoes may be acquired by using addi-

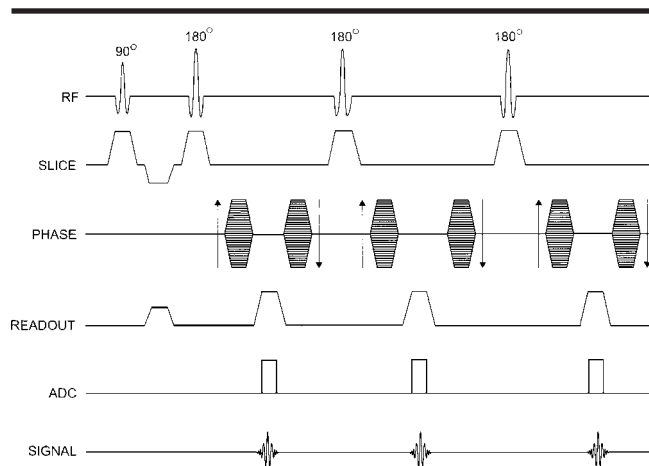


**Figure 6.** Transverse T2-weighted spin-echo image (2,000/90) shows that the cerebrospinal fluid in the lateral ventricles has high signal intensity (arrow).

tional 180° refocusing pulses and sampling periods. All TEs are measured from the common excitation pulse.

### Echo Train Spin Echo

Echo train spin echo refers to a class of techniques in which multiple spin echoes, referred to as an echo train, are used to produce the image. They are derived from the rapid acquisition with relaxation enhancement, or RARE, approach for imaging (17). The defining feature of echo train spin-echo imaging is that multiple spin echoes are acquired per excitation pulse with each echo having a different amount of phase encoding (Fig 7). Amplitude variations between the echoes are a result of both the phase-encoding amplitude and the TE for that echo. The TE is referred to as an effective TE to indicate the presence of multiple TE echoes producing each image. It is the TE at which the echoes primarily responsible for image contrast are measured (also known as the center of k space). The number of echoes acquired from each excitation pulse is known as the echo train length or turbo factor, whereas the time between each echo is known as the echo spacing. The major advantage of echo train spin echo is that the number of excitation pulses necessary to acquire the raw dataset is reduced, enabling a substantial reduction in measurement time:  $T_{\text{echo train}} = \text{TR} \times (\text{no. of acqui-}$



**Figure 7.** Echo train spin-echo pulse sequence timing diagram. Illustrated is an echo train length of three. Multiple refocusing RF pulses are used to produce multiple echoes. Each echo is detected in the presence of a constant amplitude readout gradient, following a unique amplitude of the variable phase-encoding gradient. Arrows indicate the stepping direction of the phase-encoding tables. ADC = analog-to-digital conversion.

tions)  $\times$  [(no. of phase-encoding steps)/ (echo train length)].

A distinctive feature of echo train spin-echo sequences is that the signal intensity of fat is greater than that observed on conventional spin-echo images obtained with comparable parameters. In addition, magnetic susceptibility difference artifacts are lessened due to the reduction of  $T_2^*$  effects by the multiple repetition of  $180^\circ$  refocusing RF pulses and gradients.

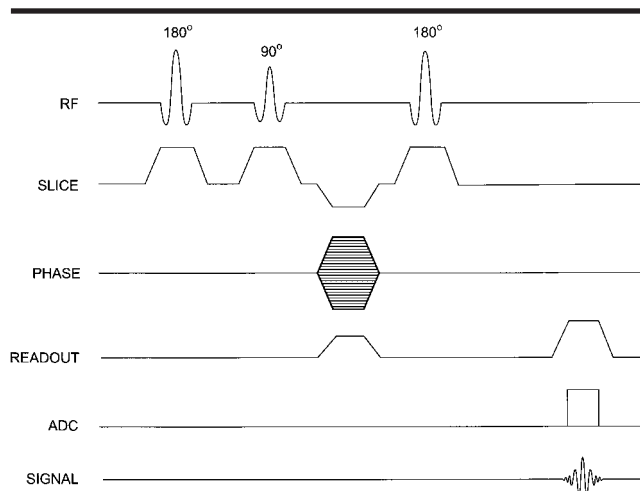
## INVERSION RECOVERY

Inversion-recovery sequences are typically derived from spin-echo sequences. A  $180^\circ$  RF pulse is applied prior to the primary excitation pulse to invert the net magnetization (Fig 8). Following this pulse,  $T_1$  relaxation occurs in which the net magnetization from each tissue passes from an inverted condition through zero net magnetization to a relaxed condition (Fig 9). A user-definable inversion time (TI) determines how much time is allowed for  $T_1$  relaxation. Subsequent excitation pulses are used to produce transverse magnetization for the resultant localization process. Inversion-recovery images have considerable  $T_1$  weighting and provide excellent contrast control through choice of the TI. The choice of TE also determines the amount of  $T_2$  weighting. Table 2 lists some of the common inversion-recovery techniques available.

The major applications of inversion-recovery sequences are for suppression of tissue. Selecting a TI when a tissue is at

zero net magnetization will cause the tissue to generate no signal. The TI time for this condition, known as the null point or null time for the tissue, is 0.692 times the  $T_1$  for the tissue to be suppressed, assuming TR is much longer than the tissue  $T_1$ . Inversion-recovery sequences are often used for signal suppression in two tissues: (a) short TI inversion recovery (STIR) technique for fat, and (b) fluid-attenuated inversion recovery (FLAIR) technique for cerebrospinal fluid. Clinical images using the STIR technique are characterized by low signal intensity from fat and high signal intensity from fluid-containing structures, rendering this technique very sensitive to disease processes that result in the accumulation of extracellular fluid (Fig 10). Two variations of inversion recovery for multisection imaging are used. If the TI is relatively short (such as is used for STIR imaging), then all RF pulses are applied to and signal is detected from a section before advancing to the next section. For a long TI (such as for FLAIR), another approach that is used is an interleaved or non-sequential method. In this approach, all inversion pulses are applied in order, then the excitation and refocusing pulses are applied for each section.

One of the major limitations of inversion-recovery sequences is that they require a long TR. This is necessary so that the net magnetization is as close to full relaxation as possible prior to the inversion pulse. The resultant long measurement times limit the usefulness of the technique for routine  $T_1$ -weighted imag-



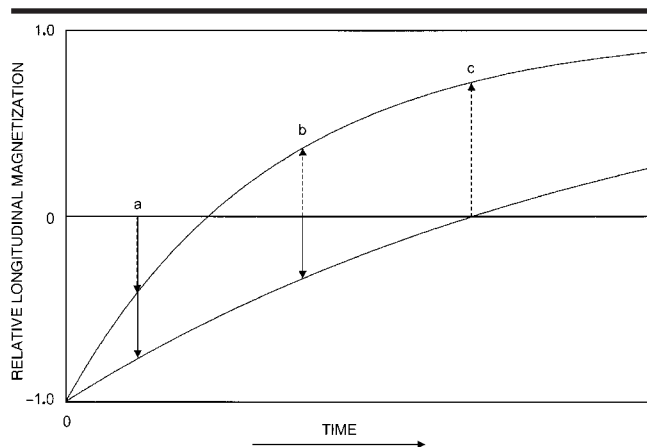
**Figure 8.** Inversion-recovery pulse sequence timing diagram. A  $180^\circ$  inversion pulse is applied prior to a  $90^\circ$  excitation pulse of a spin-echo acquisition. ADC = analog-to-digital conversion.

ing. With the advent of echo train spin-echo techniques, inversion-recovery pulses can be applied to produce echo train inversion-recovery techniques. Echo train inversion-recovery techniques combine the contrast control of standard inversion recovery with the measurement time reduction of the echo train spin-echo techniques. Currently, most applications of inversion recovery use an echo train inversion-recovery technique.

Another complication in inversion-recovery sequences is the fact that inverted net magnetization will produce a signal of opposite phase from more relaxed net magnetization. Images that take this phase into account are known as phase-sensitive images. Phase-sensitive images have unambiguous pixel values defined for tissues based on the TI, but have background air with moderate pixel values. Images that ignore the phase but consider only the magnitude of the signal are known as magnitude or modulus images. Magnitude images have low pixel values for air, but tissues with short and long  $T_1$  values relative to the TI are assigned similar pixel values, leading to an uncertainty in the tissue  $T_1$  (Fig 9 b).

## GRADIENT ECHO

A major class of pulse sequences is known as gradient-echo sequences (18,19). While spin-echo sequences use a  $180^\circ$  RF pulse to refocus the transverse magnetization, gradient-echo sequences are characterized by the lack of a  $180^\circ$  refocusing RF pulse. Instead, the echo signal is formed following a reversal of gradient pulses



**Figure 9.** T1 recovery curves following 180° inversion pulse. *a*, Tissue with a short T1 value (dashed arrow) and that with a long T1 value (solid arrow) have negative longitudinal magnetization and are assigned different pixel values. *b*, Tissue with a short T1 value (dashed arrow) has positive longitudinal magnetization, and tissue with a long T1 value (solid arrow) has negative longitudinal magnetization. Phase sensitive image reconstruction assigns different pixel values. Magnitude image reconstruction assigns equal pixel values. *c*, Tissue with a short T1 value (dashed arrow) has positive longitudinal magnetization, and tissue with a long T1 value has zero longitudinal magnetization and contributes no signal intensity to the image.



**Figure 10.** Sagittal echo train inversion-recovery, STIR image (5,000/30/150 [TR msec/TE msec/TI msec]) shows that fat has low signal intensity, such as that for bone marrow, and fluid has high signal intensity, such as that for suprapatellar effusion (arrow).

only (Fig 11). There are several consequences of this feature. Gradient-echo sequences enable high-contrast images to be acquired with use of a short TR (<300 msec). The image contrast is also much more dependent on the excitation pulse angle. The section loop will be shorter, enabling more sections to be acquired per TR when a multisection loop is used. Shorter TEs are possible than with spin-echo-based sequences. The absence of the 180° pulse reduces the RF power deposited in the patient, alleviating problems with potential tissue heating. Gradient-echo sequences are more sensitive to sources of constant dephasing such as field inhomogeneity or differences in magnetic susceptibility. This susceptibility sensitivity often produces image artifacts but may also be exploited to improve tissue contrast, such as in studies of reticuloendothelial system cells with use of iron-oxide-based contrast agents (eg, ferumoxides) or in hemorrhage due to the paramagnetic nature of blood breakdown byproducts. These additional sources of dephasing make the TE in gradient-echo sequences sensitive to T2\* effects rather than just the spin-spin relaxation time. Two-dimensional multisection, 3D volume, and sequential section loop modes are used. Table 3 lists some of the common gradient-echo sequences currently in use.

The lack of the 180° refocusing pulse also makes the relative contributions from

fat and water in gradient-echo sequences dependent on TE. Hydrogen in fat and water have a difference in resonant frequencies of 3.5 parts per million (ppm) due to the difference in molecular structures and will dephase relative to each other following an excitation pulse. In spin-echo sequences, this dephasing is reversed by the 180° refocusing pulse so that fat and water are always in phase when the signal is detected. For gradient-echo sequences, the phase difference between fat and water hydrogen in the image depends on the TE. In-phase gradient-echo images are acquired by using TE values for which the fat and water hydrogen have the same relative phase. Out-of-phase images are acquired by using TE values for which the fat and water hydrogen have the opposite phase. In-phase and out-of-phase gradient-echo images are frequently used to assess the relative fat content in a lesion (Fig 12).

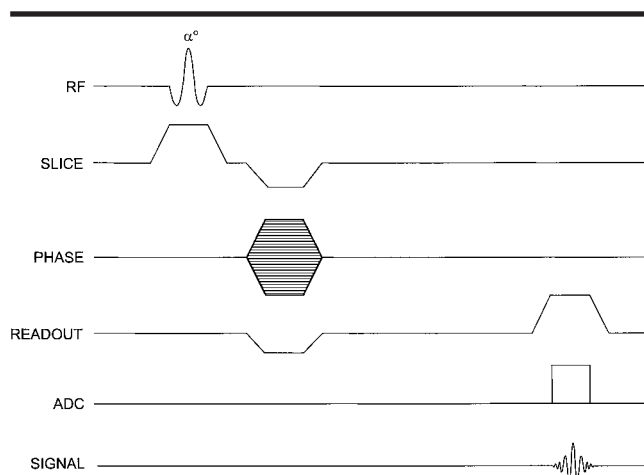
### Spoiled Gradient Echo

There are numerous variations on the basic pulse scheme of gradient-echo sequences that affect the intrinsic contrast of the image. One major categorization is the presence or absence of residual transverse magnetization at the time of the next excitation pulse. One class of gradient-echo sequences uses spoiling after signal detection to minimize residual transverse magnetization (20). This spoil-

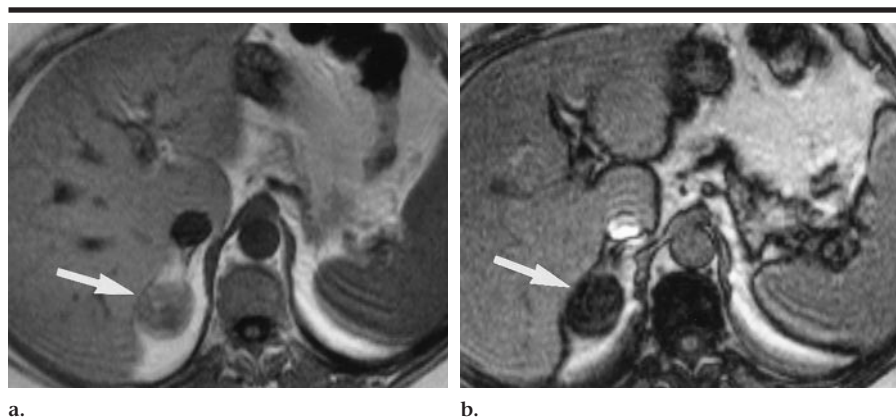
ing may be done by applying gradient pulses to “spoil” or “crush” the transverse magnetization or by “randomly” varying the phase of the RF excitation pulse each application, termed RF spoiling. In either case, any residual transverse magnetization is reduced to a minimum so that only longitudinal magnetization remains at the time of the subsequent excitation pulse. Spoiled gradient-echo techniques enable acquisition of T1-weighted or intermediate-weighted images and are particularly useful for breath-hold studies following administration of a contrast agent in abdominal imaging (Fig 13).

### Refocused Gradient Echo

Another class of gradient-echo sequences does not use spoiling, but retains the residual transverse magnetization for as long as possible. This group is known as refocused gradient-echo sequences or steady-state sequences and is most often used to produce T2\*-weighted images. The amount of T2\* weighting depends on the TR, TE, and the flip angle (also referred to as “excitation angle”). For a long TR and low flip angles, the transverse magnetization decays rapidly through intrinsic T2\* relaxation processes. If a short TE is used, the image contrast is intermediate-weighted and is comparable to that



**Figure 11.** Gradient-echo pulse sequence timing diagram. This class of sequences is characterized by the absence of a 180° refocusing pulse. Echo formation is accomplished by application of gradient pulses of opposite polarity (readout direction). ADC = analog-to-digital conversion.



**Figure 12.** Transverse (a) in-phase (140/4.5, 80° flip angle) and (b) out-of-phase (140/2.25, 80° flip angle) spoiled gradient-echo images. Increased fat content in adrenal gland (arrow) due to adrenal adenoma causes a reduction of signal intensity in b compared to that in a.

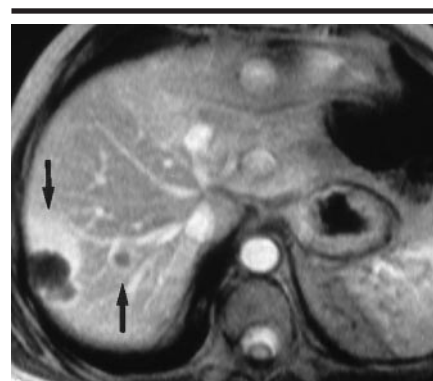
obtained by using spoiled gradient-echo sequences acquired with the same measurement parameters. Substantial T2 weighting can be obtained by using short TR and high flip angles (in actuality, the weighting is dependent on  $T1 \div T2$ ). In this case, the residual transverse magnetization is present at the time of the next excitation pulse. Care must be used in the selection of TR and flip angle to balance the large transverse magnetization necessary for T2 and the amount of T1 saturation effects.

There are several examples of refocused gradient-echo sequences. The most common type acquires the echo from net magnetization formed following the excitation pulse. However, as multiple RF pulses are applied, the maintenance of transverse magnetization enables addi-

tional echoes to be formed that can be used in the formation of images. In particular, spin echoes begin to form that are refocused at the time of the subsequent RF pulse (Fig 14). Images formed from these echoes are T2-weighted but have different spin polarity characteristics from that of the post-RF echoes. Under the proper circumstances, the pre- and postexcitation pulse signals can be combined to form images that have substantial T2 weighting with minimal T2\* dephasing effects.

### Magnetization-prepared Gradient Echo

In many instances, the choice of TR, TE, and flip angle does not provide sufficient contrast between tissues. An ex-

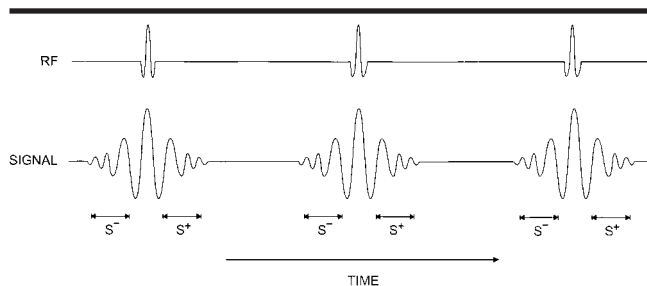


**Figure 13.** Transverse spoiled gradient-echo image (130/4, 80° flip angle) acquired following administration of a T1 relaxation contrast agent shows metastatic liver lesions (arrows) from colon cancer.

ample of this occurs when very short TRs are used to reduce the measurement time. A method to provide additional contrast is to apply an additional RF pulse or pulses to “prepare” the net magnetization in a particular fashion. The data collection period follows the preparation pulse using a low amplitude excitation pulse as the net magnetization recovers, so that the system is in a non-steady-state condition. The image contrast is determined by the time between the preparation pulse and the measurement of the low-amplitude phase-encoding steps (center of k space). The most common usage of magnetization-prepared gradient echo is for T1-weighted imaging, in which the preparation pulse is a 180° inversion pulse (Fig 15). The unique T1 relaxation time causes each tissue to contribute different amounts of signal intensity to the image (Fig 16). T1-weighted magnetization-prepared gradient-echo sequences performed by using a non-section-selective 180° inversion RF pulse are characterized by strong T1 weighting (ie, very low signal intensity from fluid-containing structures), negligible artifacts related to patient motion, and minimal phase-encoding artifacts from blood flow in vessels (Fig 17). T2-type weighting is achieved by using a 90°–180°–90° pulse train for preparation, a so-called driven equilibrium approach. Magnetization-prepared techniques are very sensitive to choices of measurement parameters and the order of raw data collection.

### HYBRID GRADIENT-ECHO-SPIN-ECHO SEQUENCES

Another variation on echo train spin echo combines multiple spin echoes and gradi-



**Figure 14.** A series of equally spaced RF pulses produces spin echoes that form at the time of subsequent echoes. Once a steady state is reached, signal is produced prior to the excitation pulse, which is echo reformation, and following the excitation pulse, which is a combination of echo and free induction decay. Images can be produced from either the preexcitation signal ( $S^-$ ) or from the postexcitation signal ( $S^+$ ).

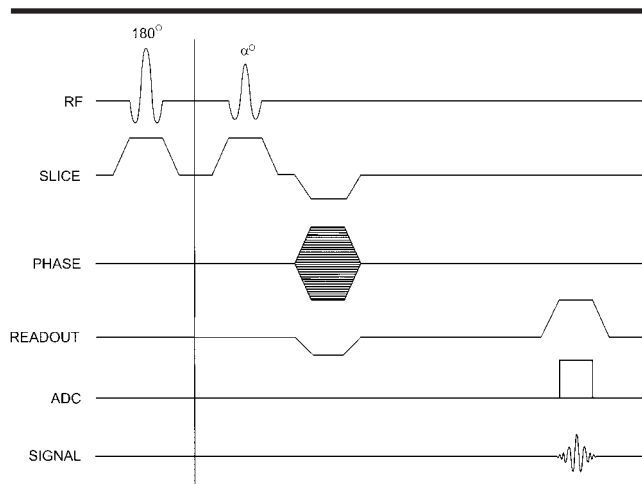
ent echoes, with each echo acquired following a different phase-encoding gradient pulse amplitude (Fig 18) (21). Normal implementations of this technique, known as hybrid gradient echo-spin echo, have the gradient echoes symmetrically placed around each spin echo, with two or four gradient echoes acquired per spin echo. Used primarily for T2-weighted imaging, hybrid gradient-echo-spin-echo sequences have features of both echo train spin echo and gradient echo. The measurement time can be substantially shortened, depending on the echo train length. The RF power deposition is reduced compared to a comparable echo train spin echo due to the use of gradient echoes. T2\* effects are also more pronounced due to the use of gradient echoes. This is manifest as a greater sensitivity to magnetic field inhomogeneities, as well as an improved visualization of hemorrhage due to magnetic susceptibility differences.

## ECHO-PLANAR IMAGING

Echo-planar imaging sequences have been recently implemented on commercially available MR systems. Echo-planar sequences are characterized by a series of rapid gradient reversals by the readout gradient, each of which produces a gradient echo (Fig 19) (22). Each echo is acquired with a different amplitude of phase encoding, so that multiple lines of raw data are acquired following each excitation pulse. Single-shot echo-planar imaging acquires all lines of raw data following a single excitation pulse and is a sequential-section technique. Multishot or segmented echo-planar imaging acquires the raw data following two or more excitation pulses and may be used in

sequential or multisection approaches. Phase encoding is performed in three fashions. One approach uses small amplitude pulses known as "blip" pulses to change the phase encoding from one echo to the next. Another method uses a continuous phase-encoding gradient throughout the entire measurement. The third approach, known as spiral imaging, varies the phase-encoding gradient in an oscillating fashion similar to the readout gradient.

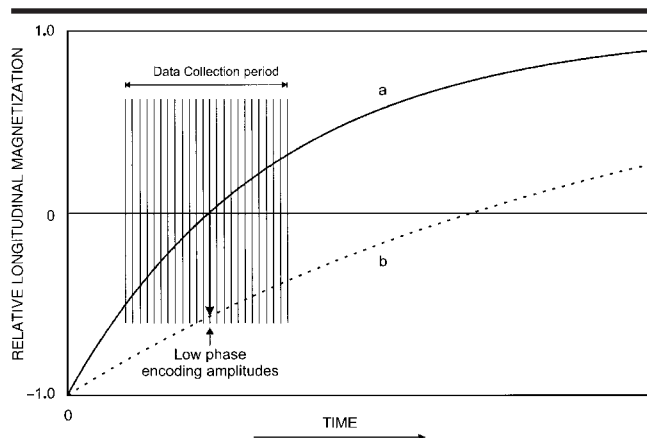
Contrast in echo-planar imaging sequences is controlled by using preparation pulses prior to data collection. The TE is considered to be an effective TE just as in echo train spin-echo imaging, since multiple echoes with different TEs are used to create each image. T1 weighting is induced by using a 180° pulse prior to the excitation pulse. Increased T2 weighting is provided by using 90°–180° pulse pairs to form a spin echo at the desired TE. Diffusion-weighted echo-planar imaging sequences use additional gradient pulses to increase the sensitivity to random molecular motion and are frequently used in the evaluation of stroke (Fig 20). Gradient-echo-type echo-planar imaging is achieved by using a single excitation pulse prior to the readout process. However, because of the gradient-echo nature of the detection technique, all echo-planar imaging techniques are sensitive to T2\* effects, such as field homogeneity distortions. Table 4 summarizes the various echo-planar imaging techniques.



**Figure 15.** Magnetization-prepared gradient-echo pulse sequence timing diagram. Preparation period is to the left of the vertical line and is executed one time. The preparation pulse illustrated is a 180° inversion pulse. Portion of the diagram to the right of the vertical line represents the data collection period and is executed multiple times, depending on the acquisition parameters. ADC = analog-to-digital conversion.

## MR SPECTROSCOPY

MR spectroscopy is a method by which metabolic information may be measured from patients in a noninvasive fashion. Whereas MR spectroscopy has been used to examine tissue samples in vitro for many years, in vivo MR spectroscopic studies of patients have only recently become a practical part of a routine clinical examination. Although many nuclei have been studied in vivo by using MR spectroscopic techniques, the most common clinical implementation of MR spectroscopy is to examine hydrogen nuclei. There are several differences between hydrogen MR spectroscopy and MR imaging studies. First, MR spectroscopic studies enable examination of other chemical species besides fat and water. In fact, the water and fat signals are usually suppressed as their signals are 1,000 times greater than the metabolites of interest. Studies of brain tissue typically examine *N*-acetylaspartate, creatine or phosphocreatine, and choline. Variations in the relative concentrations of these metabolites have provided insight into disease processes and other metabolic disorders. Second, the detection of the MR spectroscopic signal is normally made in the absence of an external magnetic field gradient. All spatial localization is performed prior to signal detection. Third, the normal display of MR spectroscopic results is a line trace known as a spectrum, though low-spatial-resolution metabolite images can be created from the



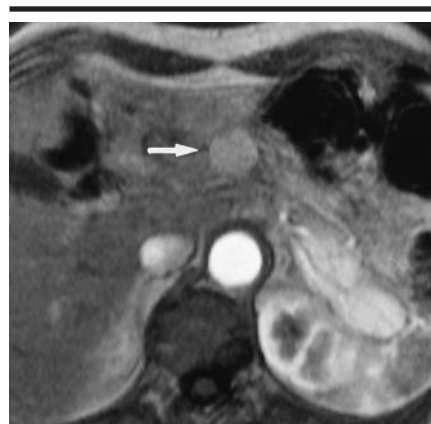
**Figure 16.** T1 recovery curve for the pulse sequence illustrated in Figure 10. The data collection period corresponds to the right side of Figure 10. At the time during the data collection when the low-amplitude phase-encoding echoes are acquired, tissue *a* (solid line) contributes no signal, while tissue *b* (dashed line) contributes considerable signal to the echo. The image will have high signal intensity from tissue *b* but low signal intensity from tissue *a*.

results as well. Fourth, the volume of tissue under examination is substantially larger than that for MR imaging studies. On a practical basis, the minimum volume of tissue from which data are acquired is  $15 \times 15 \times 15$  mm for an MR spectroscopic study compared to  $5.0 \times 0.8 \times 0.8$  mm for a typical MR imaging study. Finally, magnetic field homogeneity is more crucial in MR spectroscopy than in MR imaging. In many MR imaging applications, optimization of the magnetic field with the patient inside the magnet is not necessary if the base homogeneity is uniform. For MR spectroscopy, the magnetic field is optimized or “shimmed” to the volume of interest. This improves both the signal-to-noise ratio and the signal resolution in the resultant spectrum. More comprehensive information regarding the basic principles and clinical applications of MR spectroscopy is described elsewhere (23,24).

There are two approaches used to classify MR spectroscopic techniques. One refers to the sequence of RF pulses used to excite the volume of tissue. In general, MR spectroscopy uses three section-selective RF pulses to localize the RF energy to the volume of tissue. Two excitation schemes are used; they differ in the amplitudes of the RF pulses that are used. One approach, known as point-resolved spectroscopy (PRESS) or simply spin echo, uses a  $90^\circ$ – $180^\circ$ – $180^\circ$  pulse train and detects the spin echo following the second  $180^\circ$  pulse (25). The other technique, known as stimulated-echo acquisition

mode (STEAM), uses a  $90^\circ$ – $90^\circ$ – $90^\circ$  pulse train and acquires the stimulated echo following the last  $90^\circ$  pulse (26). Because of the timing of the particular echoes, STEAM techniques enable the use of a shorter TE compared to PRESS. The resultant spectra are very similar between the two methods, except that STEAM techniques have a substantially lower intrinsic signal-to-noise ratio than do the PRESS techniques at the same TE.

The other classification scheme refers to the method of spatial localization. Two approaches are commonly used: single voxel and multiple voxel. Single-voxel studies acquire spectra from a single volume of tissue in a measurement. The most common techniques use perpendicular section-selective RF pulses to localize the energy to the desired volume by using either the PRESS or STEAM pulse scheme (Fig 21). Multiple-voxel studies acquire spectra from several voxels during a single measurement. The most common implementation of this is known as chemical shift imaging. The primary approach for spatial localization in this method uses gradient tables analogous to those in MR imaging to phase encode the spectra according to the position within the section (27). The most common implementation is known as 2D chemical shift imaging or MR spectroscopic imaging, in which one or more narrow sections of tissue are excited (Fig 22). A double phase-encoding process divides each section into smaller regions enabling spectra to be measured from multiple volumes of tissue within each sec-



**Figure 17.** Transverse T1-weighted magnetization-prepared image (8/4.3/163,  $8^\circ$  flip angle). One second of imaging produced this image with minimal artifact from respiratory motion and slight artifact (arrow) from blood flow in the aorta.

tion. Both PRESS and STEAM excitation methods are used with both single-voxel studies and 2D chemical shift imaging techniques. Table 5 summarizes the various spectroscopic techniques.

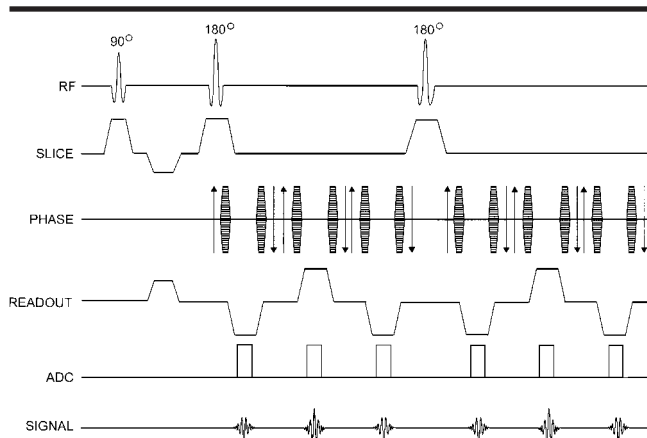
## TECHNICAL MODIFICATIONS

In addition to the heretofore described pulse sequences, additional combinations of RF and gradient pulses are often added to produce additional tissue contrast or to remove artifactual signal intensity. The resultant signal intensity depends on the timing and amplitude relationships between the various pulses. Table 6 lists some common modifications that are made to pulse sequences.

### Spatial Presaturation

Spatial presaturation pulses are used to suppress undesired signals from tissues within the imaging volume based on their location. Common examples include the suppression of signal from peristaltic motion of bowel in spinal imaging and the removal of flow artifacts from the aorta and vena cava in spinal and abdominal imaging. For spinal imaging, large anterior abdominal presaturation regions are used, while for abdominal imaging, superior and inferior regions saturate the blood before it enters the imaging volume.

Spatial presaturation pulses are frequency-selective RF pulses applied in conjunction with a gradient pulse that excites the defined tissue volume. These pulses are typically applied prior to the



**Figure 18.** Hybrid gradient-echo-spin-echo pulse sequence timing diagram. Illustrated is an echo train length of six. Arrows indicate the stepping direction of the gradient tables. ADC = analog-to-digital conversion.

imaging section pulses during sequence execution. They may be applied once per section excitation or once per TR period. Owing to their rapid occurrence, the presaturation pulses saturate the selected tissue so that its steady-state net magnetization is much smaller than the net magnetization for the remaining tissue of the section. The rate of presaturation pulse application relative to the TR determines the amount of signal difference of the saturated tissue relative to the unsaturated tissue. In addition, spoiler gradients are applied to dephase any transverse magnetization following the presaturation pulse. The result is that the signal from the presaturated region is substantially less than the signal from the nonpresaturated tissue.

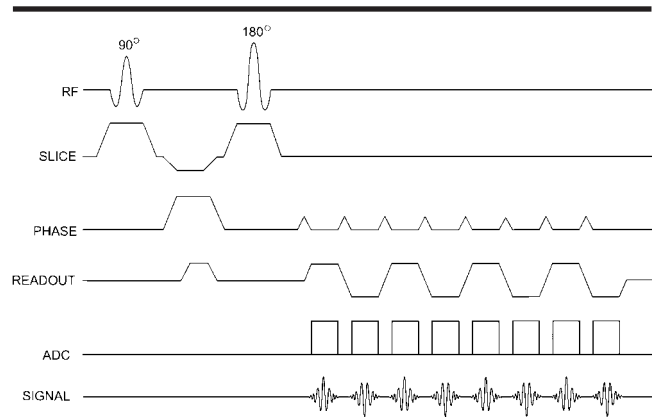
Spatial presaturation pulses are a useful and easily implemented method for reducing signal from undesired tissue. Spatial presaturation pulses may not remove all signal from the selected tissue. The saturated tissue experiences T1 relaxation during the time between the presaturation pulse and the imaging excitation pulse so that some longitudinal magnetization is present within the presaturated region at the time the section-excitation pulse is applied. Signal may be generated from the saturated region that may have substantial amplitude, depending on the particular TR for the measurement and the tissue T1 values. For this reason, presaturation pulse angles greater than 90° are often used to compensate for this relaxation. In addition, spatial presaturation pulses increase the amount of RF energy that is deposited in the patient. This may limit the number of sections or presaturation pulses that may be acquired

without exceeding safe levels of RF power deposition, as determined by the specific absorption rate, or SAR, monitor.

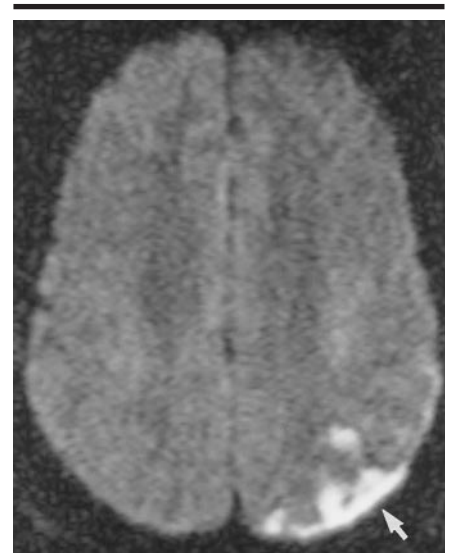
### Fat Suppression

Normal MR imaging methods depict hydrogen nuclei from both water and fat molecules within the tissue. In certain instances, it may be desirable to minimize the fat signal relative to the water signal. This is known as fat suppression, and several approaches are used to accomplish this. One approach, STIR, was described earlier in which an inversion-recovery sequence is used. The TI is chosen so that minimal fat signal intensity is measured from the section. While STIR provides excellent fat suppression, there are several drawbacks to this technique. First, a long TR is required to enable all tissues to recover their longitudinal magnetization prior to the inversion pulse, which lengthens the measurement time. Second, signal from all tissues is affected by the choice of TI. Tissues with T1 values near that of fat will also be suppressed. The TI necessary for fat suppression may reduce the contrast between other tissues. Finally, caution must be exercised following administration of a T1 contrast agent. As the agent is accumulated in various tissues, the T1 of the tissue water becomes similar to that of fat and will be reduced in signal, a behavior opposite that found in standard contrast material-enhanced T1-weighted imaging. The degree of T1 reduction is also variable as it depends on the concentration of contrast agent in a particular tissue, which in general is not predictable.

Another method to selectively visual-

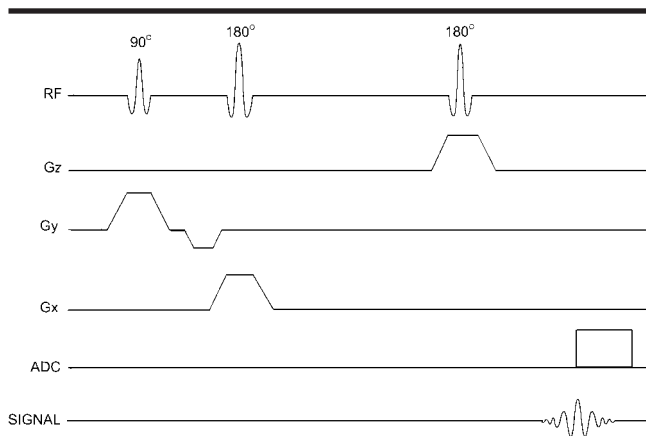


**Figure 19.** Echo-planar pulse sequence timing diagram, spin-echo type. Illustrated is an echo train length of eight. ADC = analog-to-digital conversion.

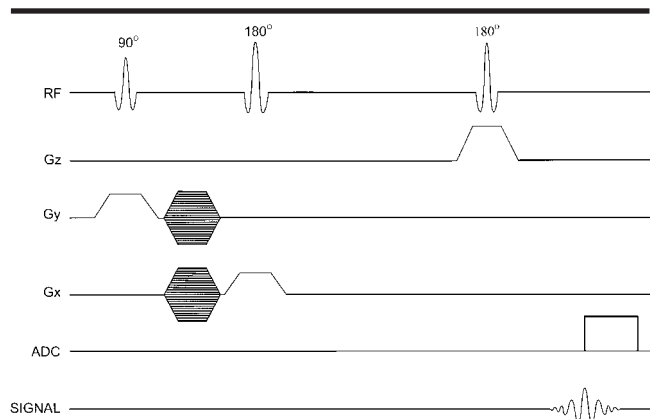


**Figure 20.** Transverse diffusion-weighted echo-planar image (TE = 123 msec;  $b$  value = 1,000 sec/mm<sup>2</sup>) shows an area of increased signal intensity (arrow) in the posterior parietal cortex corresponding to tissue with restricted motion of cellular water, indicative of stroke.

ize only the tissue water is known as fat saturation. As mentioned earlier, hydrogen nuclei in fat and water have a difference of approximately 3.5 ppm to their resonant frequencies, known as a chemical shift difference. Fat saturation applies a narrow bandwidth RF pulse centered at the fat resonant frequency in the absence of a gradient. The resultant transverse magnetization is then dephased by spoiler gradients. A standard imaging sequence may then be performed; this produces images that are based predominantly on water hydrogen nuclei. The fat-saturation pulse, often termed a chemical shift selective, or CHESS, pulse, is typically applied prior to every section excitation pulse or



**Figure 21.** Single-voxel PRESS pulse sequence timing diagram. Note that the analog-to-digital conversion (ADC) sampling begins at the peak of the echo signal.



**Figure 22.** Chemical shift imaging PRESS pulse sequence timing diagram. The compensation gradient in the  $G_y$  direction is incorporated into the gradient table. Note that the analog-to-digital conversion (ADC) sampling begins at the peak of the echo signal.

once every phase-encoding step. This pulse is very similar to a spatial presaturation pulse described in the preceding section, except that no gradient is applied during the RF excitation pulse. The signal suppression mechanism is also similar to that of spatial presaturation, in that minimal net magnetization from the fat is present at the time of the excitation pulse for the section.

Fat saturation has two main advantages over STIR imaging for fat suppression. It may be used with any type of imaging sequence. T1-weighted fat-saturation sequences may also be used with T1 relaxation agents since the agent shortens the T1 relaxation times of only the water hydrogen nuclei. The T1 reduction enables the enhanced tissues to generate substantial signal, while the fat signal remains minimal in the presence or absence of the contrast agent. However, four potential problems are inherent with fat saturation. First, there will be magnetization transfer suppression (see next section) of the water hydrogen nuclei by the saturation pulse. Second, the hydrogen nuclei of fat undergo T1 relaxation during the time between the saturation pulse and the imaging pulses and thereby contribute to the detected signal. As the fat signal approaches the water signal in amplitude, the contrast between the fat and water tissues is reduced. Saturation pulse angles greater than  $90^\circ$  can compensate for the fat relaxation to some degree. Third, fat saturation is particularly sensitive to magnetic field homogeneity. The exact frequencies for fat and water hydrogen within a voxel depend on the magnetic field that the voxel experiences. If the field homogeneity is not uniform throughout the imaging volume, the cen-

ter frequency of the saturation pulse will be off-resonance for some of the fat and will not suppress it. In some cases, the water hydrogen may be saturated rather than the fat hydrogen. For this reason, optimization of the field homogeneity to the specific patient prior to applying a fat-saturation pulse is advisable. Finally, the fat-saturation pulse increases the RF power deposition to the patient.

The use of frequency selective excitation pulses as in fat saturation can be extended to the suppression of other tissues. For some applications, suppression of the water signal is desired. This can be accomplished by choosing the center frequency of the saturation pulse to excite the water hydrogen rather than the fat. Imaging of silicone implants in breast imaging can be aided by the suppression of silicone-based hydrogen by using a saturation pulse at that hydrogen frequency.

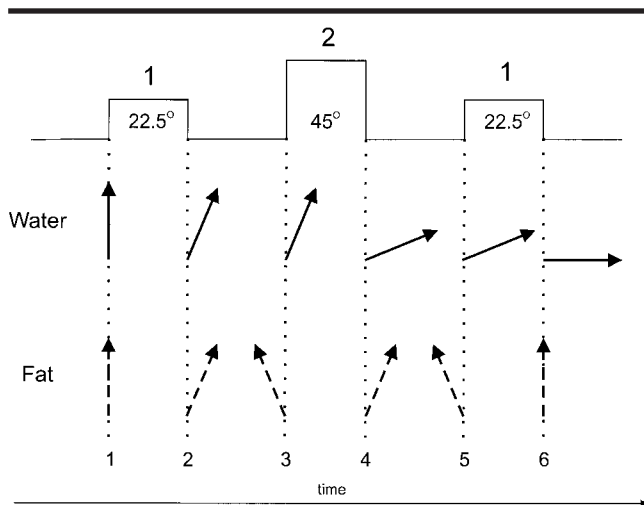
A third approach for fat suppression is known as water excitation. In this method, RF pulse combinations known as composite pulses are used to excite only the water hydrogen and leave the fat hydrogen unexcited (Fig 23). Gradient spoiling is not necessary to remove the fat signal. Because of their nature, composite pulses are very sensitive to field homogeneity. Poor homogeneity will cause undesired excitation and/or incomplete suppression. As with fat saturation, optimization of the field homogeneity to each patient is advisable.

A final technique for fat suppression is known as the Dixon method (28,29). It is also known as fat-water imaging or chemical shift imaging, though it is not a spectroscopic technique as is 2D chemical shift imaging described earlier. The Dixon method is based on a spin-echo pulse sequence. Two or three sets of im-

ages are acquired at each section position. One set is a normal spin-echo sequence. The other set(s) is acquired with the readout refocusing gradient adjusted to vary the relative amounts of water and fat signal in each voxel. The subsequent combination of the measured images allows the production of images containing only water or fat. Good field homogeneity is critical to ensure that each image has consistent behavior of fat and water within each voxel of the section. Optimization of the field homogeneity to each patient is also advisable for this technique.

### Magnetization Transfer Suppression

Another suppression technique similar in hardware implementation to fat saturation is magnetization transfer suppression. Magnetization transfer suppression uses a narrow bandwidth saturation pulse centered 1–10 kHz away from the central water frequency applied in the absence of a gradient (30). This pulse selectively excites and saturates the hydrogen adsorbed to macromolecules within tissues, known as bound water. Following the pulse, an exchange occurs between the bound water and the unsaturated mobile water, transferring the saturation to the mobile water. This causes a loss of steady-state magnetization and reduces the signal from the mobile water. This is the process of magnetization transfer suppression. Contrast is enhanced between tissues that undergo magnetization transfer (water-containing tissues) and those that do not (fat-containing tissues). Magnetization transfer pulses are used in spin-echo or gradient-echo sequences to produce additional signal suppression of tissue water. They also contribute to the



**Figure 23.** Water excitation. A 121 (1, 2, 1 at top of image) composite pulse is shown with a total flip angle of  $90^\circ$ . Prior to the first RF pulse (1 at bottom of image), both water (solid arrows) and fat (dashed arrows) hydrogens are unexcited. At the end of the first RF pulse (2 at bottom of image), both are excited  $22.5^\circ$ . Because of the difference in resonant frequencies between fat and water, the fat hydrogen becomes out of phase with the water hydrogen. The time for the second RF pulse (3) is chosen so that the fat hydrogen is exactly  $180^\circ$  out of phase. At the end of the second RF pulse (4), the water proton is rotated  $67.5^\circ$  while the fat hydrogen is rotated  $-22.5^\circ$ . A similar delay is chosen between the second and third RF pulses (5). At the end of the third RF pulse (6), the fat hydrogen is at  $0^\circ$  (unexcited), while the water hydrogen is rotated  $90^\circ$ .

total amount of RF energy that is deposited in the patient and may cause specific absorption rate limitations for the measurement.

Magnetization transfer suppression is most often used for suppression of normal-tissue water in studies in which reduction of the signal from normal tissue is desired. One example is the use of magnetization transfer in 3D time-of-flight MR angiography. In time-of-flight MR angiography, contrast between normal, stationary tissue and flowing blood is provided through T1 saturation of stationary-tissue water by repeated excitation pulses. Additional suppression of the stationary-tissue water with use of magnetization transfer pulses enables smaller vessels to be visualized (Fig 24). Another application of magnetization transfer is T1 studies following the administration of a contrast agent. T1 contrast agents shorten the T1 relaxation time for tissues in which the agent is located. Comparison of images acquired before and after administration of contrast material enable determination of dispersal of the agent. Use of a magnetization transfer pulse during the postcontrast measurement reduces the signal from the nonenhanced tissues, providing greater contrast with the contrast-enhanced tissue. Magnetization transfer

pulses also aid in the examination of white matter diseases, such as multiple sclerosis, by enhancing the conspicuity of white matter lesions.

### MOTION ARTIFACT REDUCTION TECHNIQUES

The most problematic aspect of MR imaging is its sensitivity to moving tissue. Motion during MR imaging can produce severe image artifacts in the phase-encoding direction. The severity of the artifact depends on the nature of the motion, the time during data collection when the motion occurs, and the particular pulse sequence and measurement parameters. The most critical portion of the data collection period for artifact generation is during the collection of echoes following low-amplitude phase-encoding steps (center of k space). Motion during the high-amplitude phase-encoding steps causes blurring but not severe signal misregistration on the image. Three approaches are commonly used to reduce the severity of the motion artifact on the final image. Two of these approaches, triggering and respiratory compensation, acquire information when the moving tissue is in a reproducibly stationary position. The third tech-



a.



b.

**Figure 24.** Effect of magnetization transfer pulse. (a) No magnetization transfer pulse, (b) magnetization transfer pulse. (40/7,  $25^\circ$  flip angle, for a and b.) Normal gray matter and white matter are suppressed by application of a magnetization transfer pulse, enabling smaller diameter vessels (arrow) to be seen.

nique, gradient motion compensation, correctly acquires the moving and stationary tissue so that there is no misregistration. While none of these approaches will completely remove motion artifacts from the image, use of one or more of these techniques will substantially reduce the effect of motion on the final images.

### Triggering or Gating

Prospective triggering or gating initiates the measurement following a peri-

odic signal produced by the patient such as a pulse or heartbeat. The signal detection for a particular section always occurs at the same time following the timing signal. Since the moving tissue is in the same relative position at this time, there is minimal misregistration of signal and considerable reduction of the resultant motion artifact. An important application of this technique is heart imaging, in which data collection is synchronized to the electrocardiographic signal of the patient. The peak R wave is normally used as the timing reference point. Each phase-encoding step is acquired at the same point in time following the R wave so that the heart is in the same relative position. Since the measurement time per section is shorter than the duration of the R-R interval, many images can be acquired within one heartbeat. Two approaches are commonly used to subdivide the images. For spin-echo imaging, each image is acquired at a different section position and a different time point in the cardiac cycle. These images are typically T1-weighted (depending on the R-R time interval) with minimal blood signal and are used for morphologic studies of the heart. Another approach has considerable blood signal and is known as cine heart imaging. In this method, multiple images are acquired at each section position during the R-R time interval. Rapid display of these images will allow a dynamic visualization of the heart during the different phases of the cardiac cycle.

There are several potential problems with triggered studies. One is that the measurement time is usually longer for a triggered study than for the corresponding nontriggered study. Time must be allowed from the end of data collection to the next trigger signal to ensure that the trigger signal is properly detected by the measurement hardware. This extra time is typically 150–200 msec to allow for variation of the heart rate of the patient. The total measurement time will be extended by approximately 1–2 minutes. Another problem with heart imaging occurs if the heart rate is irregular. The effective TR for a phase-encoding step depends on the R-R time interval. Variation in the heart rate causes variation in the amount of T1 relaxation for the section from measurement to measurement of each phase-encoding step. This will produce amplitude variations to the measured signal, resulting in misregistration artifacts on the final images even if detection of the triggering signal is accurate. Stability of the heart rate is most critical

during collection of the echoes following low-amplitude phase-encoding gradients. Finally, proper detection of the electrocardiographic signal from the patient is critical. Improper electrode placement may result in substantial distortion of the electrocardiographic signal by the blood during its flow through the aortic arch. In addition, the transmitted RF energy and the gradient pulses may interact with the lead wires, inducing substantial noise to the detected electrocardiographic signal. Use of high-resistance lead wires may result in heating of the wires through this coupling if the wires are arranged in a loop and may seriously burn the patient.

An alternative to prospective triggering for cine heart imaging is known as retrospective gating. In this approach, the electrocardiographic signal is measured during imaging but the data collection is not controlled by the timing signal. Instead, the data are measured in a nontriggered fashion, and the time following the R wave when each phase-encoding step was measured is stored with it. Following completion of the data collection, images are reconstructed corresponding to various time points within the cardiac cycle. The data for any phase-encoding step not directly measured are interpolated from the measured values.

Gating of the measurement can also be used for abdominal imaging to reduce artifacts from respiratory motion. Table 7 lists the various methods that are used for this. Data collection is synchronized to the respiratory cycle of the patient. Detection of the respiratory motion can be done mechanically by using a pressure transducer positioned on the abdomen or chest or through the use of navigator echoes in which an MR signal is monitored for signal fluctuations caused by chest or abdominal movement. Simple respiratory gating acquires the data when there is minimal motion, such as during end expiration. It suffers from greatly increased measurement times since the time during respiratory motion is not used for data collection. An alternative method for T1-weighted imaging, respiratory compensation, rearranges the phase-encoding gradient table so that adjacent phase-encoding steps are acquired when the abdomen is in the same relative position. Typically, low-amplitude phase-encoding steps are acquired during end expiration, while high-amplitude phase-encoding steps are acquired during end inspiration. Respiratory compensation has the advantage that the measurement time is not increased. Greatly reduced respira-

tory-induced ghosts can be achieved by means of either technique as long as there is a uniform respiration rate during imaging.

### Gradient Motion Compensation

A final method for reducing motion artifacts uses additional gradient pulses to correct for phase shifts experienced by the moving tissue (31). This technique has a variety of names, which are listed in Table 7, but all are based on the same basic principle. As described earlier, a gradient echo is generated by the application of two gradient pulses of equal magnitude but opposite polarity. Proper dephasing and rephasing of the hydrogen nuclei and correct frequency mapping of the tissue occurs as long as there is no motion during the gradient pulses. Movement during either gradient pulse results in incomplete phase cancellation or a net phase accumulation at the end of the second gradient pulse time. The amount of phase accumulation is related to the velocity of the motion. This phase accumulation produces signal intensity variations that are manifest as motion artifacts in the phase-encoding direction.

If the motion of the tissue is relatively simple with respect to time, then the induced phase changes can be predicted and may be corrected by applying additional gradient pulses. These pulses will be applied in the direction for which compensation is desired. The number, duration, amplitude, and timing of the pulses can be defined so that tissue moving with constant velocity (first-order motion), acceleration (second-order motion), and pulsatility (third-order motion) can be properly mapped within the image. For gradient-echo sequences, first-order compensation is normally sufficient for proper registration of cerebrospinal fluid, whereas for spin-echo sequences higher order compensation can often be achieved with minimal difficulty.

Gradient motion compensation results in certain restrictions for the pulse sequence. Since additional gradient pulses are applied during the section loop, the minimum TE for the sequence must be extended to allow time for their application. Higher degrees of compensation require more gradient pulses. In pulse sequences in which short TEs are desired, higher amplitude gradient pulses of shorter duration may be used. This will limit the minimum field of view that may be used for the sequence. In practice, only modest increases in TE and the minimum field of view are normally required.

**TABLE 1**  
**Spin-Echo Sequences**

Manufacturer	Single-Echo Spin Echo	Multiple-Echo Spin Echo	Echo Train Spin Echo	Hybrid Gradient Echo-Spin Echo
Siemens Medical Systems (Erlangen, Germany)	Single spin echo	Spin echo Double echo	Turbo spin echo (TSE) Half Fourier acquisition turbo spin echo (HASTE)	Turbo gradient spin echo (TGSE)
GE Medical Systems (Milwaukee, Wis)	Spin echo	Multiecho multiplanar (MEMP) Variable echo multiplanar (VEMP)	Fast spin echo (FSE) Single shot FSE (SS-FSE)	Gradient-spin echo (GRASE)
Philips Medical Systems (Best, the Netherlands)	Spin echo Modified spin echo	Multiple spin echo (MSE)	Turbo spin echo (TSE) Ultrafast spin echo (UFSE)	GRASE
Picker International (Highland Hts, Ohio)	Spin-echo throughput heightened rapid increased flip T2 (THRIFT) Phase symmetrized rapid increased flip spin echo (PRISE)	Multiecho THRIFT	Fast spin echo (FSE)	GRASE

**TABLE 2**  
**Inversion-Recovery Sequences**

Manufacturer	Standard Inversion Recovery	Echo Train Inversion Recovery	Interleaved Excitation	Magnitude Reconstruction	Phase Sensitive Reconstruction
Siemens Medical Systems	IR	TurboIR	Interleaved	Absolute value Magnitude (standard)*	TrueIR
GE Medical Systems	Multiplanar IR (MPIR)	Fast multiplanar IR (FMPIR)	Nonsequential	Modulus (standard)*	...
Philips Medical Systems	IR	IR-turbo spin echo (IR-TSE)	...		Real
Picker International	IR	IR	Slice interleaved		...

\* Indicates default mode.

**TABLE 3**  
**Gradient-Echo Sequences**

Manufacturer	Spoiled	Refocused, Postexcitation	Refocused, Preexcitation	Magnetization Prepared
Siemens Medical Systems	Fast low angle shot (FLASH)	Fast imaging with steady-state precession (FISP)	Reversed FISP (PSIF)	TurboFLASH Magnetization-prepared rapid acquisition gradient echo (MP-RAGE)
GE Medical Systems	Spoiled GRASS (SPGR) Fast spoiled GRASS (FSPGR) Multiplanar spoiled GRASS (MPSPGR) Fast multiplanar spoiled GRASS (FMPSPGR)	Gradient acquisition in the steady state (GRASS) Fast GRASS Multiplanar GRASS (MPGR) Fast Multiplanar GRASS (FMPGR)	Steady state free precession (SSFP)	IR-prepared fast GRASS Driven equilibrium (DE)-prepared fast GRASS
Philips Medical Systems	T1 contrast-enhanced fast field echo (T1 CE-FFE)	Fast field echo (FFE)	T2 contrast-enhanced FFE (T2 CE-FFE)	Turbo field echo (TFE)
Picker International	Radio-frequency spoiled Fourier-acquired steady state (RF-FAST)	FAST	Contrast-enhanced FAST (CE-FAST)	Rapid acquisition magnetization-prepared FAST (RAM-FAST)

## MR ANGIOGRAPHY

MR angiography is a technique for the visualization of MR signal from flowing blood located in the vascular network. The most common approaches are "bright blood" techniques, in which the data are acquired so that blood is assigned the largest pixel value in the image. Both arterial and venous blood can be studied,

either simultaneously or individually, by using spatial presaturation pulses to suppress the unwanted blood. Both 2D and 3D gradient-echo techniques are used, depending on the flow velocity and volume of tissue under examination.

There are two major classes of MR angiographic techniques that are used. Time-of-flight techniques are saturation techniques that use short TR and large

flip angles to suppress the stationary tissue in the imaging volume (32). Flowing blood enters the imaging volume and produces greater signal intensity than that of the surrounding stationary tissue. Magnetization transfer pulses are often included to provide additional signal suppression of the stationary tissue. Time-of-flight techniques are very time efficient in that the images are acquired with

**TABLE 4**  
**Echo-Planar Imaging Terminology**

Term	Synonyms	Characteristics
Single shot	...	All echoes for image acquired following one excitation pulse
Multishot	Segmented	Echoes for image are acquired following more than one excitation pulse
Blipped phase encoding	...	Small amplitude steps applied between readout gradient reversals
Constant phase encoding	...	Phase-encoding gradient is on continuously
Spiral phase encoding	...	Phase-encoding gradient has alternately positive and negative polarity

Note.—Terminology is the same for all the manufacturers.

**TABLE 5**  
**Hydrogen MR Spectroscopy Terminology**

Term	Synonyms	Characteristics
Point-resolved spectroscopy	PRESS, SE	Volume excitation using section-selective 90°-180°-180° pulse train
Stimulated echo acquisition mode	STEAM	Volume excitation using section-selective 90°-90°-90° pulse train
Single-voxel spectroscopy	SVS	Small volume of tissue excited, single spectrum produced
Two-dimensional chemical shift imaging, 2D MR spectroscopic imaging	2D-CSI, 2D-MRSI	Large volume of tissue excited, multiple spectra produced

Note.—Terminology is the same for all the manufacturers.

substantial contrast between the stationary and flowing tissue. Only one set of images at a given section position is required to visualize flow through the section. However, time-of-flight techniques do not allow for the quantification of flow velocity.

Time-of-flight techniques use either 2D sequential section or 3D volume modes of acquisition. The 2D sequential section mode is best when imaging slow flow or large anatomic regions but suffer from poor section profiles and degraded spatial resolution. Pronounced vessel misregistration can occur if there is patient motion during the measurement. Three-dimensional time-of-flight techniques have superior spatial resolution and reduced sensitivity to patient motion, but also have considerable signal loss from saturation of in-plane flow for large excitation volumes. Three approaches are used to address this problem. One technique uses multiple overlapping 3D volumes of small thickness in a sequential mode of acquisition. This technique, known as multiple overlapping thin-slab acquisition, or MOTSA, reduces the amount of in-plane

**TABLE 6**  
**Technical Modifications**

Manufacturer	Spatial Presaturation	Fat Suppression			Magnetization Transfer	MR Angiography Nonuniform Excitation
		STIR*	Fat Saturation	Water Excitation		
Siemens Medical Systems	Presaturation (Presat)	STIR	Fatsat	Water excitation	MT	Tilted optimized non-uniform excitation (TONE)
GE Medical Systems	Saturation (Sat)	STIR	Fatsat Chemsat Spectral presaturation	...	MT	Ramped excitation
Philips Medical Systems	Regional saturation technique (REST)	STIR	Spectral presaturation with inversion recovery (SPIR)	Water excitation	Magnetization transfer contrast (MTC)	TONE
Picker International	Presaturation (PS)	STIR	Chemical shift imaging—Dixon method	Water excitation	MT	Ramped excitation

\* STIR = short inversion time inversion recovery.

**TABLE 7**  
**Motion Compensation Techniques**

Manufacturer	Gradient Moment Nulling		Respiratory Motion
	Name of Technique	Nature of Compensation	
Siemens Medical Systems	Gradient motion rephasing (GMR)	First order readout, section, and phase directions	Respiratory gating
GE Medical Systems	Flow compensation (FLOW COMP)	First order readout and section directions	Respiratory gating
Philips Medical Systems	Flow adjustable gradients (FLAG)	First and second order readout, section, and phase directions	Respiratory compensation (Exorcist)
Picker International	Motion artifact suppression technique (MAST)	First, second, and third order readout and section directions	Respiratory triggering Phase-encoding artifact reduction (PEAR)
			Respiratory gating Respiratory ordered phase encoding (ROPE)

Note.—Not all combinations are possible in all directions for all sequences.

flow saturation compared to a standard 3D technique covering the same total imaging volume. Patient motion between acquisitions of the 3D volumes is still problematic. A second approach uses contrast media to shorten the T1 relaxation time of the blood. This method has been successfully used in conjunction with 3D volume acquisitions in the evaluation of renal artery stenosis and in the examination of peripheral flow in extremities (33). Spatial presaturation pulses used to suppress undesired flow are ineffective with this technique due to the considerable shortening of the blood T1 relaxation time. The final approach uses a primary excitation pulse with a nonuniform or ramped RF excitation profile. Blood at the proximal side of the imaging volume is excited with a greater flip angle relative to the distal side, thereby producing higher signal intensity on the images. Acronyms for this technique are summarized in Table 6.

The other class of MR angiographic techniques is known as phase-contrast techniques. Phase-contrast MR angiography is a subtraction technique in which two measurements are performed at each section position to provide visualization of the flowing tissue (34). In one measurement, gradient pulses are used to produce a phase shift for flowing tissue. The amount of phase shift depends on the flow velocity and the gradient pulse amplitude. The second measurement may acquire images by using a different gradient amplitude to produce a different phase shift or may acquire images with flow compensation only. Subtraction of images from the two measurements results in high signal intensity from flowing tissue and minimal signal intensity from the stationary tissue. The gradient amplitudes are defined to produce the maximal amount of phase shift that can be accurately visualized, which corresponds to the flow velocity that produces the maximum signal intensity. Phase-contrast techniques provide excellent suppression of background tissue and enable quantification of the flow velocity and direction.

However, because of the multiple measurements involved, phase-contrast techniques will be longer than time-of-flight techniques. Knowledge of the expected maximal flow velocity prior to the measurement is also necessary to ensure optimal visualization.

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