CHAPTER 8

Signal suppression techniques

Chapter 6 presented the concept of a pulse sequence and described several classes of pulse sequences. The RF and gradient pulses were applied in very precisely defined ways to uniformly affect the signal intensity from all the protons within the volume of measured tissue. Additional RF excitation pulses may be added to any of these sequences to manipulate the net magnetization $\boldsymbol{M}$ of some of the tissue within the imaging volume and differentially affect its contribution to the detected signal. One approach uses frequency-selective saturation pulses, either applied in conjunction with a gradient (spatial presaturation) or in its absence (fat/water saturation, magnetization transfer suppression). Another approach uses the chemical shift frequency difference inherent in tissues to change the relative phase of the signal contribution. This is the basis for water/fat excitation using composite RF pulses and the Dixon method for fat/water suppression. In all of these cases, an increase in the minimal *TR* for the sequence (sequence kernel time) is required to implement the pulses. In addition, the additional RF pulse(s) increases the total RF power deposition to the patient. Limitations due to the specific rate of RF energy absorption (SAR) (see Chapter 14) may be required, particularly for spin echo-based sequences using short *TR*.

8.1 Spatial presaturation

Spatial presaturation pulses are frequency-selective RF pulses applied in conjunction with a gradient pulse based on their location. They have center frequencies and gradient amplitudes different from the pulses used for the imaging volume (Figure 8.1). Spatial presaturation pulses are used to suppress undesired signals from locations within the imaging volume. They are often employed to suppress an artifactual signal from peristaltic and respiratory motion in lumbar spine imaging. They are also used to reduce blood flow artifacts from the aorta or inferior vena cava in abdominal imaging by saturating the blood before it enters the imaging volume. These types of pulses are also used to produce a saturation tag for analyzing the direction of blood flow or cardiac motion.

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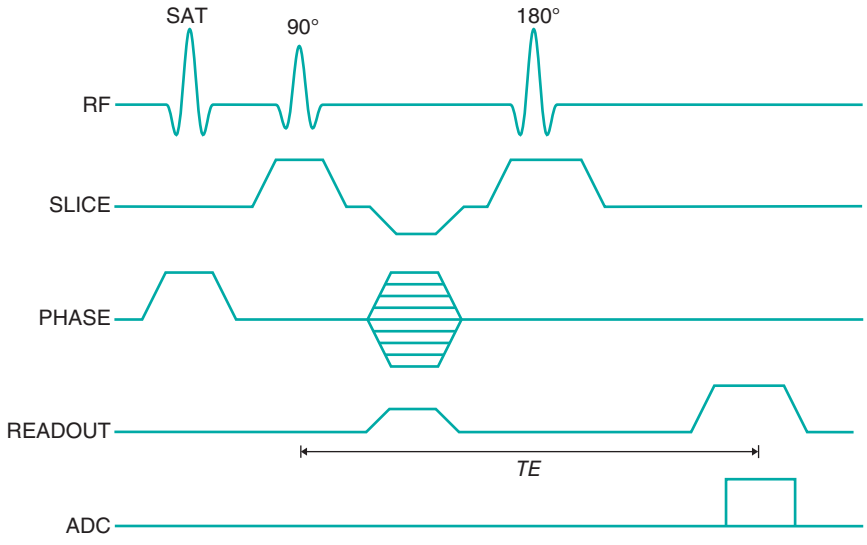


Figure 8.1 Standard single-echo spin echo sequence timing diagram including a spatial presaturation pulse. The saturation pulse (labeled SAT) is applied prior to the primary slice excitation pulse (labeled 90°). The RF pulse center frequency and bandwidth and the gradient amplitudes for the presaturation pulse are independent of these variables for the slice excitation pulses.

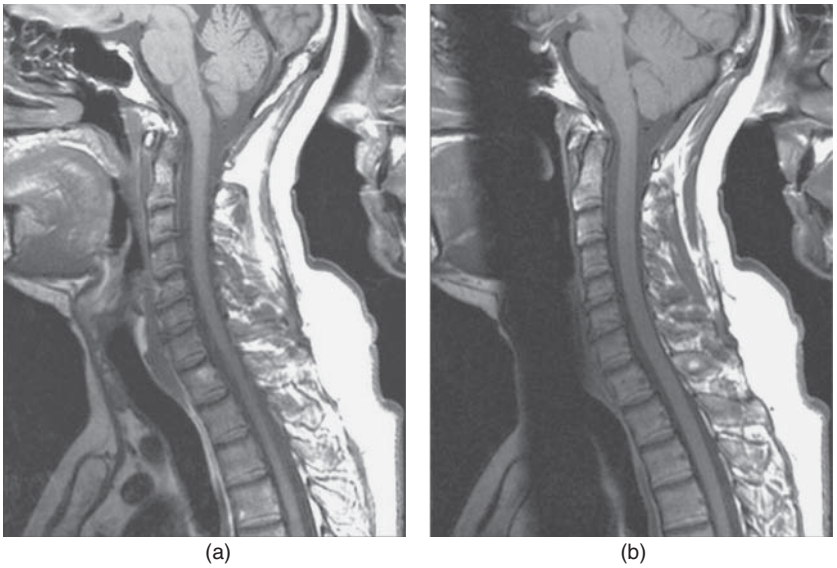


Figure 8.2 Application of a spatial presaturation pulse to moving tissue will suppress signal from that tissue. Measurement parameters are: pulse sequence, spin echo; *TR*, 500 ms; *TE*, 16 ms; excitation angle, 90° ; acquisition matrix, N_{PE}, 192 and N_{RO}, 256; FOV, 210 mm PE × 280 mm RO; N_{SA}, 3; slice thickness, 4 mm. (a) No presaturation pulse; (b) coronal spatial presaturation pulse, suppressing artifact from swallowing.

Spatial presaturation pulses are usually applied prior to the imaging slice pulses during sequence execution. They may be applied once per slice loop or once per *TR* time period. Due 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 slice. In addition, spoiler gradients are applied to dephase any residual transverse magnetization following the presaturation pulse. The result is that the signal measured from the presaturated region is significantly less than the signal from the nonpresaturated tissue (Figure 8.2). In addition to the problems regarding sequence kernel times and RF power deposition previously mentioned, spatial presaturation pulses will not remove all signals from the selected region. Tissues in the saturated region experience *T1* relaxation during the time between the presaturation pulse and the imaging excitation pulse so that longitudinal magnetization is present within the presaturated region at the time the slice excitation pulse is applied. This generates a signal from the saturated region that may have significant amplitude, depending on the particular *TR* for the measurement and the tissue *T1* values. The amount of apparent signal suppression depends on the amount of signal produced in the saturated region relative to the signal produced in the unsaturated region.

8.2 Magnetization transfer suppression

Another signal suppression technique similar in implementation to spatial presaturation is magnetization transfer suppression. A frequency-selective RF pulse is used, but in the absence of a gradient, to indirectly saturate tissue water (Figure 8.3). Water within a tissue is either mobile (freely moving) or bound (adsorbed to macromolecules). The bound fraction water protons have a very short *T2* value due to the rapid dephasing they undergo. The resonance peak for these spins is very broad and normally does not contribute significantly to the measured signal. The mobile water molecules have a much longer *T2* and a narrow resonance peak. These two resonances are superimposed at the same center frequency (Figure 8.4).

Magnetization transfer suppression is accomplished using a narrow bandwidth saturation pulse known as a magnetization transfer (MT) pulse centered 1–10 kHz away from the central water frequency applied in the absence of a gradient. Because of their broad resonance peak width, only the bound water protons are irradiated by the RF pulse and become saturated. An exchange occurs between the bound water protons and the unsaturated mobile water protons that transfers the saturation to the mobile fraction protons, causing a loss of steady-state magnetization and reducing the signal from the mobile fraction protons. This process is called 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.

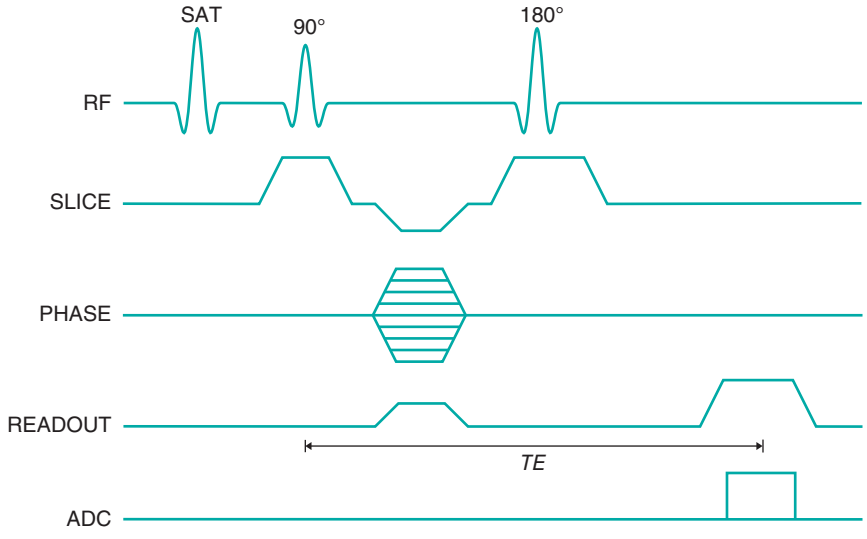


Figure 8.3 Standard single-echo spin echo sequence timing diagram, including a frequency-selective presaturation pulse. The saturation pulse (labeled SAT) is applied prior to the primary slice excitation pulse (labeled 90°). The RF pulse center frequency and bandwidth for the presaturation pulse are independent of these variables for the slice excitation pulses. Note the absence of the associated gradient pulse for the presaturation pulse compared to Figure 8.1.

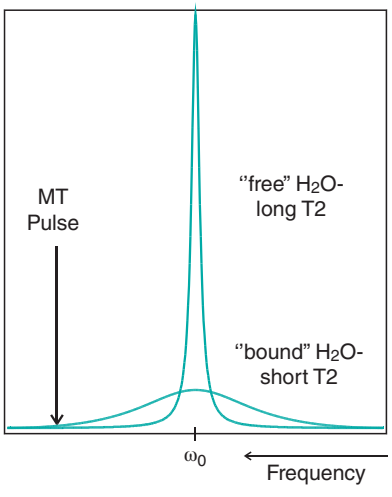


Figure 8.4 Magnetization transfer suppression. Mobile or "free" tissue water has protons with long *T2* values and produces a narrow resonance peak. Water adsorbed or "bound" to macromolecules has protons with short *T2* values and produces a wide resonance peak normally not visualized in an image. Both types of water protons have the same resonant frequency. The magnetization transfer RF pulse is applied at a frequency different (off-resonance) from the water to saturate the bound water protons. Exchange between the bound and free water transfers the saturation to the free water protons, reducing signal intensity from the free water.

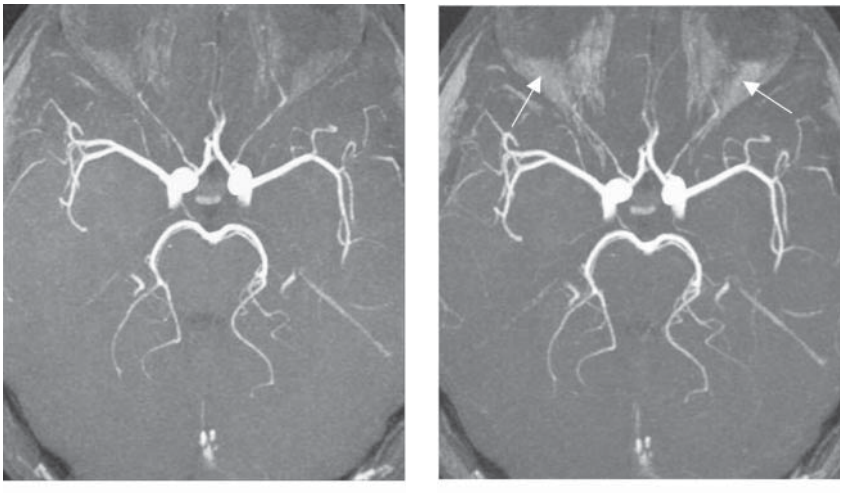


Figure 8.5 Effects of magnetization transfer in three-dimensional MR angiography. Application of MT pulse suppressed background signal from gray and white matter, enabling better visualization of blood vessels. An apparent increase in signal from suborbital fat is observed (arrows). Measurement parameters are: pulse sequence, three-dimensional refocused gradient echo, postexcitation; *TR*, 42 ms; *TE*, 7 ms; excitation angle, 25°; acquisition matrix, N_{PE}, 192 and N_{RO} 512 with twofold readout oversampling; FOV, 201 mm PE × 230 mm RO; N_{SA}, 1; effective slice thickness, 0.78 mm. (a) No MT pulse; (b) MT pulse.

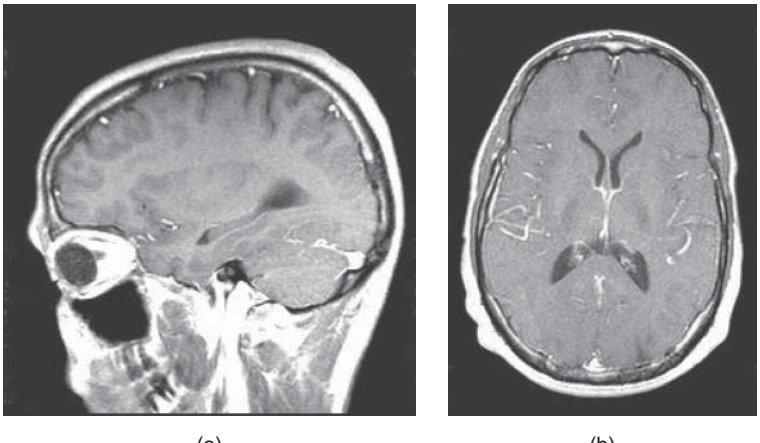


Figure 8.6 Effects of magnetization transfer in *T1*-weighted imaging following contrast administration. Application of MT pulse suppresses background signal from normal matter, enabling better visualization of contrast-enhanced tissues such as tumors or vascular structures. (a) No MT pulse; (b) MT pulse.

Magnetization transfer suppression is most often used to reduce a signal from normal tissue water in studies where this tissue is of little interest. Two examples illustrate this. Time-of-flight MR angiography (see Chapter 11) is a technique for visualizing blood flow within the vascular network. Suppression of the normal brain tissue water using magnetization transfer pulses enables smaller vessels to be distinguished (Figure 8.5). The other application of magnetization transfer is *T1* studies following the administration of a contrast agent. *T1* contrast agents shorten the *T1* relaxation time for tissues where the agent is located (see Chapter 15). Comparison of images acquired before and after contrast administration enables determination of the agent dispersal within a tissue. Use of a magnetization transfer pulse during the postcontrast measurement reduces the signal from the unenhanced tissues, increasing their contrast with the enhanced tissue (Figure 8.6).

8.3 Frequency-selective saturation

Another application for a frequency-selective saturation pulse without an accompanying gradient is to directly suppress signals from spins that are visualized in an image. The spins that are typically selected are either fat or water protons. Normal MR imaging methods visualize protons from both water and fat molecules within the tissue. As mentioned in Chapter 2, fat and water have a chemical shift difference of approximately 3.5 ppm to their resonant frequencies. Frequency-selective saturation uses a narrow bandwidth RF pulse centered at either the fat or water resonant frequency applied in the absence of a gradient (Figure 8.7). The resulting transverse magnetization is then dephased by spoiler gradients. A standard imaging sequence may then be performed, which produces images from the

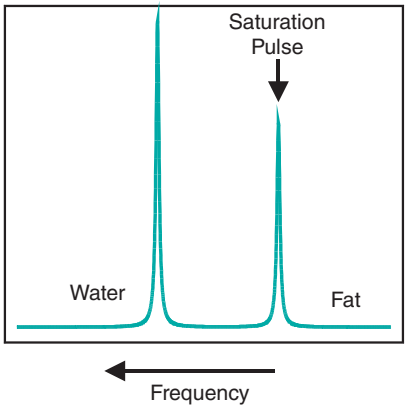
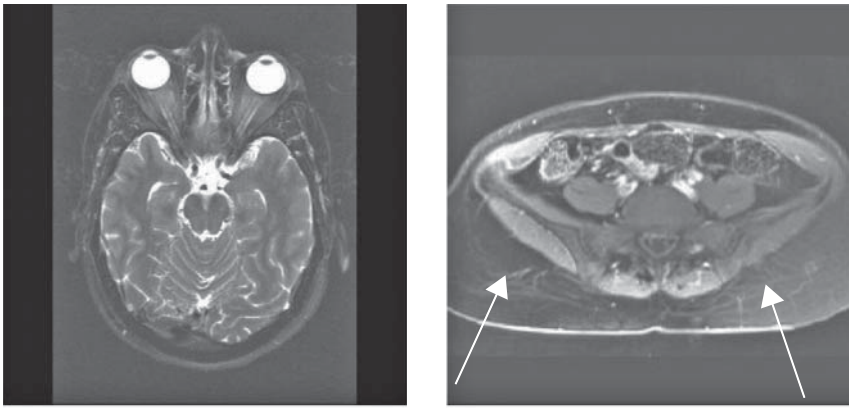


Figure 8.7 Frequency spectrum of fat and water. Fat saturation applies an additional RF excitation pulse centered at the fat resonant frequency. This pulse is applied prior to the primary slice excitation pulse, so that the signal from the slice is produced primarily from the water.



(a) (b)

Figure 8.8 Frequency-selective saturation (fatsat) pulse is applied to suppress signal from fat protons. (a) With a homogeneous magnetic field, the suppression of fat is uniform throughout the slice; (b) with a nonhomogeneous magnetic field, the saturation pulse suppresses fat well in one region of the image and poorly in another region (arrows).

other type of protons within the slice (Figure 8.8a). The number of saturation pulses applied during the sequence loop is variable, ranging from once per *TR* time period to once per slice excitation pulse. The signal suppression mechanism is similar to that of spatial presaturation described previously, in that minimal net magnetization from the saturated protons is present at the time of the excitation pulse for the slice.

While signal from either fat or water protons may be suppressed using frequency-selective presaturation, its most frequent usage is for suppression of fat. Fat saturation has two main advantages over STIR imaging (see Chapter 6) for fat suppression. It may be incorporated into virtually any type of imaging sequence. *T1* fat saturation sequences may also be used with gadolinium-based *T1* contrast agents since the contrast agent shortens the *T1* relaxation times of only the water protons (see Chapter 15). The *T1* reduction enables the enhanced tissues to generate significant signal while the fat signal remains minimal in the presence or absence of the contrast agent.

Three potential problems are inherent with frequency-selective presaturation, in addition to the problems of increased slice loop time and RF power deposition. One is that there will be magnetization transfer suppression (see above) of the water protons by the saturation pulse if fat saturation is performed. The second problem is that the saturated protons undergo *T1* relaxation during the time between the saturation pulse and the imaging pulses and will contribute to the detected signal. Multiple fat saturation pulses within a slice loop may be necessary to achieve the desired signal suppression, increasing the required *TR*.

Finally, frequency-selective presaturation is particularly sensitive to the magnetic field homogeneity. The exact resonant frequency for a fat and a water proton depend upon the magnetic field that a voxel experiences. If the homogeneity is not uniform throughout the imaging volume, the center frequency of the saturation pulse will be off-resonance for some of the spins and will not be effective in suppression (Figure 8.8b). In some cases, the water protons may be saturated rather than the fat protons. For this reason, optimization of the field homogeneity to the specific patient prior to applying a frequency-selective presaturation pulse is advisable.

8.4 Nonsaturation methods

Two methods for signal suppression rely on the inherent frequency difference between fat and water, but are not saturation-based techniques. One method uses a gradient echo sequence with a composite RF pulse (see Chapter 5) for slice/volume excitation. Known as water excitation, the pulse amplitudes and relative timings within the composite pulse are chosen to selectively excite the water protons and leave the fat protons unexcited (Figure 8.9). Composite pulses are very sensitive to field homogeneity since differences in the resonant frequency change the cycling time for fat relative to water. Poor homogeneity causes undesired excitation and/or incomplete suppression. For this reason, optimization of the field homogeneity to each patient is advisable.

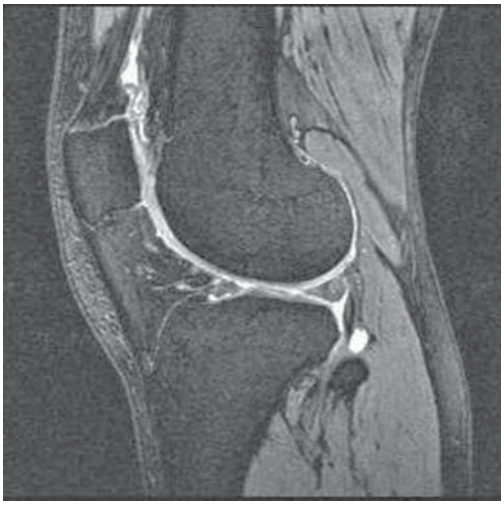
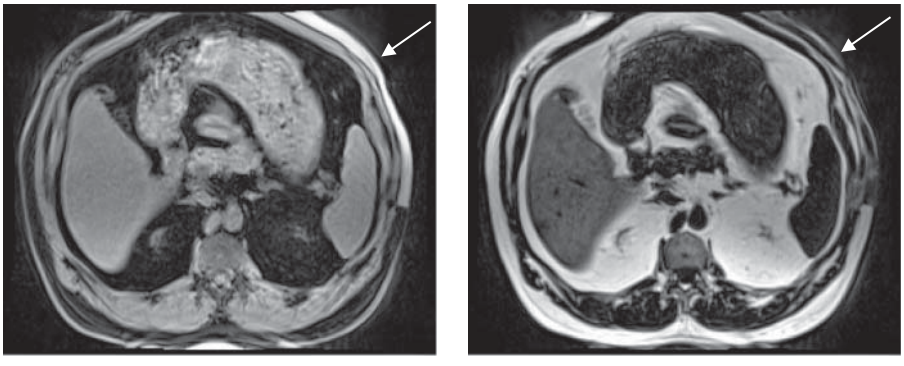


Figure 8.9 Water excitation image acquired using a composite RF pulse. Measurement parameters:pulse sequence, three-dimensional gradient echo, combination pre-and postexcitation; *TR*, 25.4 ms; *TE*, 9 ms; excitation angle, 35° cumulative; acquisition matrix, N_{PE}, 256 and N_{RO}, 256; FOV, 160 mm PE × 160 mm RO; N_{SA}, 1; effective slice thickness, 1.56 mm.



(a) (b)

Figure 8.10 Dixon method for fat suppression: (a) water image; (b) fat image. Poor magnet homogeneity results in a phase wrap, causing the fat and water to appear in the incorrect images (arrows).

Another method for fat suppression is known as the Dixon method (Dixon, 1984), which is a mathematical model-based approach and can be used both with spin echo and gradient echo scans. Recall in Chapter 6 that the spin echo pulse sequence has both an RF echo and a gradient echo that are superimposed in time during signal collection. The resulting image will have both fat and water contributing with the same signal phase. If the gradient prephase area in the readout direction is modified, then the fat echo can be made to be 180° in phase relative to the spin echo. This will reverse the relative fat contribution in the detected signal. Addition of the two images will create an image of only water, while subtraction will create an image of only fat (Figure 8.10). Subsequent refinements have made the Dixon method very insensitive to field homogeneity by accounting for differences in the resonant frequency that change the cycling time for fat relative to water. Acquisition of a third echo with a different amount of fat contribution may be necessary to avoid artifacts such as illustrated in Figure 8.10.