Techniques of Fat Suppression

Ian Cameron, PhD FCCPM

MR Physicist, The Ottawa Hospital, 501 Smyth Rd., Ottawa, ON, Canada, K1H 8L6

Introduction

The Magnetic Resonance (MR) signal used to create MR images is induced by hydrogen nuclei. All hydrogen nuclei in the object being imaged (i.e. the subject) contribute to the MR signal but for many of these nuclei (e.g. hydrogen nuclei on fatty acid chains in cell membranes) their contribution decays too quickly to contribute to the measured signal. The actual signal is composed mainly of contributions from hydrogen nuclei residing on water molecules and fat molecules in adipose tissue. In this article these hydrogen nuclei will be referred to as "water spins" and "fat spins", respectively. The net effect of these spin groups is an induced effective magnetic field or magnetization. The magnetization for these groups will be referred to here as water magnetization and fat magnetization, respectively. The signals generated by each of these magnetizations will be referred to as water and fat signals, respectively.

There are many situations in clinical MRI where it is desirable to remove the fat contribution from the total MR signal without affecting the water signal. Fat suppression techniques can be used to enhance tissue contrast and lesion conspicuity, to determine if the tissue of interest has a high or low lipid content and to remove artefacts. Specific examples include suppression of the marrow signal from around joints and in vertebrae and the suppression of the fat signal in the orbits to better differentiate tissues of interest (cartilage and ligaments, bone metastases, optic nerve, etc.) from surrounding fatty tissue. Suppression of hyperintense fat signals may also be useful in postcontrast images where lesions may appear bright. Further examples will be discussed in the other sections of this syllabus.

To suppress the fat signal for a given MR sequence a fat suppression module is typically inserted at the beginning of an otherwise normal MRI sequence. To prepare the signal such that the fat contribution is as small as possible, without perturbing the water signal, one or more of the following properties is exploited: 1) fat and water have different resonant frequencies, 2) they have different Larmor precession frequencies and 3) they have different T_1 relaxation times.

The water signal has identical contributions from its two hydrogen atoms. Fat signals can have contributions from many inequivalent hydrogen atoms (e.g. CH_3, CH_2, $CH=CH$, etc.) each of which will have distinct resonance frequencies. The most important fat resonance peaks occur at frequencies between 3.3 and 3.5 ppm lower than the water resonance peak. The signal from the CH_2 groups in the aliphatic chain gives the strongest peak and it occurs 3.35 ppm lower than water [1], which corresponds to a frequency of 214 Hz lower than the water resonance frequency at 1.5 T and 428 Hz lower at 3.0 T.

A number of quite different fat suppression techniques are available, each with its own advantages and disadvantages. The rest of this article provides of an explanation of the following fat suppression techniques: 1) Spectral Fat Saturation, 2) STIR, 3) SPIR, 4) the Dixon method and 5) water excitation.

Spectral Fat Saturation:

With this form of fat suppression the fat resonance is excited selectively and then the signal is "spoiled" using gradient pulses. The fat spins are initially tipped into the transverse plane using a special 90° pulse that affects only the fat spins. After the rf pulse, the fat spins are aligned perpendicular to the main magnetic field, B_0, while the water spins are still parallel to B_0. If a signal were to be measured at this point it would have contributions from fat spins only. However, spoiler gradient pulses are used to de-phase the fat spins causing the fat signal to decay to zero without affecting the water spins, which are still in equilibrium. At this point, the fat signal is said to be "saturated". The fat signal has been suppressed and a standard MR sequence can now be initiated. The resulting image should, in principle, have no contribution from fat spins.

In order to take advantage of the different resonant frequencies for fat and water molecules, the overlap of their spectral peaks must be as small as possible. This involves both maximizing the frequency difference between the peaks and minimizing the spread of the peaks. The main fat resonance peak is at a frequency which is lower than the water resonance frequency by 3.35 ppm [1] (i.e. 214 Hz at 1.5 t and 428 Hz at 3.0 T). As the magnetic field increases the frequency difference between the two resonances also increases. Consequently, fat suppression techniques based on this property tend to work better at higher magnetic field strengths. The width of the peaks should, in principle, be a reflection of the distribution of magnetic environments inherent to the subject. However, the observed distribution of resonance frequencies, as given by the width of the measured resonance peak, may be dominated by inhomogeneities of the applied magnetic field rather than the magnetic field distribution inherent to the subject, although both contribute. On low field systems or poorly shimmed magnets the width of the resonance peaks may be large compared to their separation resulting in substantial overlap of the fat and water resonance peaks. Under these conditions, fat saturation does not suppress the fat signal very well. Spectral fat saturation does not work well on low field systems or poorly shimmed magnets. Conversely, at high fields and/or on well shimmed magnets the fat and water resonance peaks do not overlap, leading to good fat saturation. The homogeneity of the magnetic field, and therefore the quality of the fat suppression, is normally also better near the isocentre of the magnet.

The rf pulse used in Spectral Fat Saturation must be designed to excite only fat spins. The central frequency of the rf excitation must be lower than the water resonance frequency by 3.35 ppm (214 Hz at 1.5 T and 418 Hz at 3.0 T). The frequency range excited, or spectral bandwidth (not to be confused with readout bandwidth), must also be carefully selected so that only fat spins are excited. Note that both the central frequency offset relative to the water resonance frequency and the spectral bandwidth of the fat excitation pulse change with the magnetic field strength used. The quality of the fat saturation also deteriorates if the flip angle of the rf pulse is different at different positions in the subject unless an adiabatic spectrally selective rf pulse is used. Adiabatic pulses are specifically designed to be insensitive to rf spatial nonuniformity.

For sequences with multiple repetitions (e.g. a conventional spin echo or gradient echo sequence) the fat suppression must typically be applied before each repetition, unless TR $\ll$ T_{1fat}, as may be the case with some gradient echo sequences. To compensate for regrowth of the fat signal during the sequence a tip angle greater than 90° can sometimes be used for the initial fat excitation pulse.

STIR (**S**hort **T**I **I**nversion **R**ecovery or **S**hort **T**au **I**nversion **R**ecovery):

STIR is an inversion recovery sequence where the value of the inversion time, TI, is chosen such that the fat signal does not contribute to the resulting image [2]. With this fat suppression technique the total signal (fat and water) is initially inverted (i.e. flip angle = 180°) and allowed to relax back to equilibrium via T_1 relaxation. The inversion rf pulse causes spins initially parallel to the main field to become oriented anti-parallel to the main field (i.e. the initial magnetization, $\mathbf{M_0}$, becomes $-\mathbf{M_0}$). As the spins relax back to their equilibrium configuration (with the magnetization parallel to the main field) the signal for each spin group will evolve from a negative signal, through zero (i.e. the null point), to a positive signal, at a rate which is determined by the T_1 of the spin group. Since at 1.5 T, T_{1fat} = 260 ms [3] and for most other tissues $T_1 \geq 500$ ms [3], the null point for the fat signal will occur much sooner than for other tissues. The actual T_1 values vary with magnetic field strength but this general trend is maintained. At the fat null point the fat signal will be zero but the signal for the other tissues will normally be non-zero. Therefore, if a standard MRI sequence is started when the fat signal is at its null point then the fat spins will not contribute to the resulting image.

In principle, the fat null point will be reached when TI = T_{1fat}*ln2 = (260 ms)*0.693 = 180 ms at 1.5 T [4]. In practice, the optimal value will also depend on other sequence parameter settings (e.g. TR); typically TI is chosen to be less than the theoretical null point (e.g. 150 ms at 1.5 T). TI will also depend on field strength since T_{1fat} increases with field strength.

Since STIR is an IR technique, the resulting fat suppressed image will be inherently T_1-weighted; however, the T_1-contrast will be inverted (unless signal phase is considered) relative to conventional T_1-weighting: tissues with a short T_1 will appear dark while tissues with a long T_1 will be bright. All signals will relax during the TI period so the resulting tissue signals will be smaller by the amount these tissues relax during this time. For example, if TI = 180 ms and $T_{1tissue}$ = 700 ms then the magnitude of the tissue magnetization at TI (i.e. the start of the image acquisition) will be $|-0.55M_0|$ or 55% of its full value. T_{1fat} is normally much shorter than the T_1 of other tissues but this is not always the case (e.g subacute hemorrhage). When $T_{1fat} \approx T_{1tissue}$ the signal from that tissue will also be substantially suppressed when STIR fat suppression is used. T_2-weighting can also be introduced, if desired, by using a moderate to long TE. The T_2 contrast is not inverted.

One disadvantage of STIR is its relatively long acquisition time, even with interleaved IR, since TR must be longer than for conventional T_1-weighted acquisitions to give the spins enough time to recover. STIR is also sensitive to spatial nonuniformity of the applied rf pulse (unless an adiabatic rf pulse is used). If the strength of the rf pulse varies from one position to another within the subject then the tip angle of

the inversion pulse, and the quality of the fat suppression, will also vary with position. However, STIR is insensitive to inhomogeneity of the main static magnetic field, B_0. Unlike spectral fat saturation, STIR works well even on low field systems and poorly shimmed magnets.

SPIR (**S**pectral **P**resaturation with **I**nversion **R**ecovery):

SPIR is a fat suppression technique that makes use of both selective excitation of the fat signal and T_1 relaxation. As with STIR, an inversion rf pulse is used, but unlike STIR, this pulse is designed to excite only the fat spins. This is similar to the situation with Spectral Fat Saturation except that the spectrally selective rf pulse in STIR is an inversion pulse which causes the fat magnetization to tip through 180° (i.e. to be anti-parallel to the main magnetic field) without affecting the water signal. After the inversion, the fat spins evolve back to their equilibrium orientation parallel to the main field. As they pass through the fat null point (i.e. when the fat signal is zero) a conventional MR sequence is initiated. The resulting image will be fat suppressed. The TI value at which fat is nulled will, in principle, be the same as for STIR; however, when other sequence parameters are considered the optimal setting may be different.

Although the implementation of STIR and SPIR may seem similar the resulting images are significantly different. SPIR has several important advantages over STIR. SPIR fat suppression does not cause inherent T_1-weighting since the water spins are not affected (i.e. inverted) by the fat suppression procedure. The MR sequence that follows the fat suppression module can, in principle, be any conventional MR sequence. If the acquisition parameters for this sequence are set to create T_1-weighting, the contrast is not inverted as it would be with a STIR sequence. SPIR also has inherently higher pixel intensities than STIR for tissues that do not contain fat. Furthermore, signals from tissues with T_1 values that are close to that of fat, such as contrast enhanced tissues, will not be suppressed, as they would be with STIR.

Disadvantages of SPIR are that 1) it requires good separation of the fat and water resonance peaks 2) it is sensitive to rf pulse spatial nonuniformity (i.e. flip angle changing with position) and 3) it lengthens the exam time significantly relative to a similar acquisition without fat suppression. On low field systems or poorly shimmed magnets the fat and water resonances may overlap making it impossible for the spectrally selective rf pulse to excite only fat spins. SPIR also requires that the inversion pulse be the same for all fat spins in the slice. If the intensity of the rf pulse varies spatially (i.e. poor B_1 uniformity) there will be a range of tip angles and fat suppression will be better in some parts of the image than in others.

Sometimes SPIR is implemented with an adiabatic spectrally selective pulse. When this is the case, the SPIR fat suppression module is not sensitive to B_1 nonuniformity since adiabatic pulses are specifically designed to be insensitive to B_1 nonuniformity.

The Dixon Method:

The principle upon which the 2-point Dixon technique [5,6] is based is that, since fat and water have different resonance frequencies, they will also precess in the transverse plane at different rates (i.e. they have different Larmor precession frequencies). By adjusting the sequence timing, the phase of the fat spins relative to the water spins in the transverse plane can be adjusted to whatever phase angle is desired when the signal is acquired. In the 2-point Dixon method two images are acquired; one with the fat and water spins in-phase and the other with them out-of-phase. These images can be obtained from separate acquisitions or as different echoes of the same acquisition. If these two images are added together pixel by pixel the result will be a fat suppressed image. Conversely, if these images are subtracted the resulting image will be water suppressed. Note that the complex pixel intensities must be used for these calculations. If only the pixel magnitudes are used the resulting images will be incorrect when the fat signal coming from the voxel is stronger than the water signal.

The phase difference between the water and fat spins in the transverse plane following an rf excitation pulse is given by $\Delta\theta = 2\pi\Delta f * TE$ where Δf is the difference in their resonance frequencies (in Hz) and TE is the echo time (in seconds). From this equation it can easily be shown that $\Delta\theta = n\pi$ (i.e. the in and out of phase conditions) when TE = $2.30n$ ms for n = 0, 1, 2, ... , since Δf = 217 Hz at 1.5 T. Similarly, at 3T it can be shown that $\Delta\theta = n\pi$ when TE = $1.15n$ ms for n = 0, 1, 2, ... It is necessary to have good gradient hardware at 3T in order to achieve a double gradient echo with TE's of 1.15 ms (out-of-phase) and 2.30 ms (in-phase).

The discussion given in the previous paragraph applies to gradient echo sequences. The situation with spin echo sequences is slightly more complicated since there is both an excitation (90°) pulse and a refocusing (180°) pulse. The refocusing pulse acts to undo the dephasing that occurs between the two pulses such that the fat and water spins are in-phase again at $t = 2\tau$, where τ is the time between the 90° and 180° pulses. In this case, $\Delta\theta = 2\pi\Delta f * (TE - 2\tau)$, from which it can be shown that, for a spin echo sequence, $\Delta\theta = n\pi$ when TE - 2τ = $2.30n$ ms for n = 0, ±1, ±2, ... , at 1.5 T and $\Delta\theta = n\pi$ when TE - 2τ = $1.15n$ ms for n = 0, ±1, ±2, ... at 3.0 T.

The 2-point Dixon method assumes that the phases of the fat and water spins in the transverse plane are determined exclusively by the applied magnetic field, B_0. B_0 inhomogeneity or other effects that could affect the phase (e.g. susceptibility variations and eddy currents) are ignored. Normally these effects are not important but, for cases where they cannot be neglected, the 3-point Dixon method [7] can be used. The 3-point Dixon method is very similar to the 2-point Dixon method except that an extra image is required. Typically, images with $\Delta\theta$ = 0, π, and 2π will be obtained. The first two correspond to the images used in the 2-point Dixon method and the third is used to correct for miscellaneous phase errors. The $\Delta\theta$ = 0 and 2π signals should have the same phase behaviour and any deviation from this will be due to B_0 inhomogeneity, susceptibility variations, eddy current effects, etc. The phase error can be determined from these two images and the correction applied to the $\Delta\theta$ = 0 and π images. These corrected in-phase and out-of-phase complex images are then added and/or subtracted to get fat and/or water suppressed images, respectively.

One disadvantage of the Dixon methods on some systems may be increased exam time since two (or more) images are required. However, these images can normally be acquired at the same time using a two (or more) echo sequence. Another disadvantage of the sequence is that the TE values used are normally quite short so that the images do not have T_2-contrast. A benefit of the sequence is that four image sets with quite different tissue contrast are obtained: 1) an in-phase image, 2) an out-of-phase image, 3) a water only image and 4) a fat only image. The out-of-phase images are sometimes useful, particularly in abdominal imaging, since the border between fatty and non-fatty tissues appears dark. This is due to destructive interference of out-of-phase fat and water signals in voxels at these tissue boundaries. The Dixon method is normally a very robust fat suppression technique.

Water excitation:

Fat suppression can also be done by exciting only the water spins and leaving the fat spins unaffected. This is normally achieved by using a special type of rf pulse known as a binomial pulse. In fact, binomial pulses are actually a set of rf pulses whose net effect is to produce a 90° pulse for the water spins and a 0° pulse for the fat spins. These binomial pulses are a bit longer in duration than the normal excitation pulses but since no spoiler gradient is required and there is no delay to wait for relaxation to occur, it is a rather quick way to achieve fat suppression.

The set of pulses that make up a binomial pulse are related by the binomial condition and there are many possible combinations. To illustrate how binomial pulses can be used for fat suppression, consider a binomial pulse made up of a set of three rf pulses that induce 22.5°, 45°, and 22.5° flip angles, respectively, with the time between the pulses set such that the fat spins, which are off-resonance compared to the water spins, precess about the z-axis by 180°. In this case, the net tip angle for the on resonance water spins is the sum of the three tip angles, which is 90°. The fat spins are tipped by 22.5° by the first rf pulse but by the time the second rf pulse is applied they will have precessed to the other side of the transverse plane. Thus, the net effect of the first rf pulse plus the first delay time is equivalent to a -22.5° tip angle. The second rf pulse then tips the fat spins through 45°; from -22.5° to 22.5°. Again the precession of the fat spins brings them to -22.5° and the final rf pulse, with a tip angle of 22.5°, tips the fat spins back to 0°, where they started.

Note that if the tip angles are not exactly equal to 22.5°, 45° and 22.5°, respectively, the net tip angle for the water will not be exactly equal to 90° but the tip angle for the fat spins will still be 0°, as long as the effective tip angles are in the ratio 1:2:1. Thus, fat suppression using binomial pulses is not sensitive to rf pulse spatial nonuniformity (i.e. flip angle changing with position). This technique primarily relies on having the correct interpulse delays to suppress the fat signal.

References

1. E.M. Haacke, R.W. Brown, M.R. Thompson, R. Venkatesan, Magnetic Resonance Imaging : Physical Principles and Sequence Design, New York: John Wiley & Sons, Inc. 1999. p. 422.

2. G.M. Bydder, I.R. Young, MRI: Clinical Use of the Inversion Recovery Sequence, J. Comp. Asst. Tomogr. 9:659-675, 1985.

3. M.A. Bernstein, K.F. King, X.J. Zhou, Handbook of MRI Pulse Sequences, New York, Elsevier Academic Press, 2004. p. 961.

4. M.A. Bernstein, K.F. King, X.J. Zhou, Handbook of MRI Pulse Sequences, New York, Elsevier Academic Press, 2004. p. 623.

5. W.T. Dixon, Simple Proton Spectroscopic Imaging, Radiology 153 : 189-194.

6. M.A. Bernstein, K.F. King, X.J. Zhou, Handbook of MRI Pulse Sequences, New York, Elsevier Academic Press, 2004. p. 857-887.

7. G.H. Glover, E. Schneider, Multipoint Dixon Technique for Water and Fat Proton and Susceptibility Imaging, J. Magn. Reson. Imaging 1:521-530.