

MRI Imaging Options

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As MR imaging becomes more developed, more imaging options are being made available. Currently, there exist options to reduce or eliminate artifacts, decrease scan time, and improve image contrast. A few of these options are explained briefly below.

Before the options are explained, a brief summary of the imaging parameters that affect the signal-to-noise (SNR) in MRI will be provided so that it can be understood how the imaging options affect the SNR. In the simplest terms, the SNR equation can be written as:

$$SNR \propto \text{voxel volume} \times \sqrt{\text{acquisition time}}$$

This equation can be written in terms of the imaging parameters that contribute to the *voxel volume* and the *acquisition time*. The *voxel volume* is determined by the dimensions of the pixel in the image (FOV_x/N_x and FOV_y/N_y , where FOV_x and FOV_y are the field-of-view in the phase- and frequency-encoding directions, respectively, and N_x and N_y are the number of frequency- and phase-encoding points in the image matrix, respectively) and the slice thickness (*slthick*), and can be written as:

$$\text{voxel volume} = \frac{FOV_x}{N_x} \times \frac{FOV_y}{N_y} \times \text{slthick}$$

The second term in the SNR equation, the *acquisition time*, or more specifically, the time spent acquiring data, is determined by how long the acquisition electronics are turned on during the imaging sequence. The acquisition electronics are turned on when the MR signal is being sampled. The amount of time required to sample the signal during the frequency-encoding process depends on the number of frequency-encoded samples acquired, and the amount of time spent acquiring each sample. The number of frequency-encoded samples acquired is simply equal to the number of frequency-encoding points in the imaging matrix (N_x), and the amount of time spent sampling each point is inversely proportional to the sampling rate, or the receiver bandwidth (RBW). So, the amount of time spent sampling the frequency-encoded signal is equal to $N_x \times 1/\text{RBW}$. To gather enough information to form an image, the frequency-encoded signal is sampled once for each phase encoding value (N_y). In order to improve the SNR of an image the data can be sampled more than once, and the data sets can be averaged together. If the number of averages is denoted as N_{ave} , the total amount of time spent acquiring data may be written as:

$$\text{acquisition time} = \frac{N_x}{\text{RBW}} \times N_y \times N_{ave}$$

So, in terms of scan parameters, the SNR equation can be written as:

$$SNR \propto \frac{FOV_x}{N_x} \times \frac{FOV_y}{N_y} \times \text{slthick} \times \sqrt{\frac{N_x N_y N_{ave}}{RBW}}$$

The proportionality symbol is used because the SNR is also dependent on the magnetic field strength, the imaging coil, the type of imaging sequence, the TR, TE, and tip angle of the imaging sequence, the T₁, T₂, density, and temperature of the material being imaged, etc.

Some imaging options and their effect on SNR are described below.

Rectangular Field-of-View

Selecting the rectangular field-of-view option allows acquisition and reconstruction of a field-of-view (FOV) that has a smaller extent in the phase-encoding dimension than in the frequency-encoding dimension. Reducing the FOV in the phase-encoding dimension permits a reduction in the number of phase-encoding values that must be acquired to produce the image. This leads to a reduction in the total scan time. Thus, if the extent of the object is less in one dimension than in the other, the rectangular FOV option can be used to limit the extent of the FOV in this dimension in order to achieve a shorter scan time. It is important that the object not extend outside the FOV in the reduced dimension or else it will alias, or wrap around, just as is the case for full FOV imaging.

Images acquired with full and 3/4 FOV are shown in Figs. 1a and 1b, respectively. For the image in Fig. 1a, 128 phase-encoding values were prescribed and acquired. For the image in Fig. 1b, selecting the 3/4 FOV option automatically reduced the number of phase-encoding values to 96, thereby reducing the scan time accordingly.

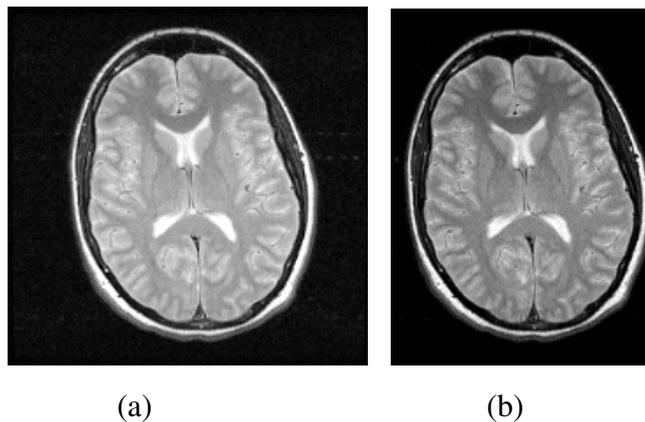


Figure 1: Images acquired with (a) full and (b) 3/4 fields-of-view. Acquisition time is proportional to the fraction of the FOV acquired.

When a rectangular FOV is imaged, the SNR is less than what it would be if the entire FOV was imaged. This results because fewer phase-encoding samples are acquired in fractional FOV images, so fewer data points contribute to producing the image. Using the SNR equation, it can be calculated that the SNR for $\frac{3}{4}$ FOV imaging is $\sqrt{\frac{3}{4}}$ times what it is for full FOV imaging. This is because the *voxel volume* remains unchanged (FOV_y and N_y are reduced by the same factor so there is no net change in the pixel size/resolution in the phase-encoding dimension), but the *acquisition time* is reduced to $\frac{3}{4}$ of what it is for full FOV imaging because the number of phase encoding values (N_y) is $\frac{3}{4}$ of what it is for full FOV imaging. Other fractions of the FOV may be imaged as well, and the SNR scales according to the square root of the fraction of the FOV imaged.

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When partial Fourier acquisition is performed in the phase encoding direction, not all of the phase-encoding values are acquired. For example, for a $\frac{3}{4} N_y$ Fourier acquisition only $\frac{3}{4}$ of the phase-encoding values are sampled, and for a $\frac{1}{2} N_y$ Fourier acquisition only half of the phase-encoding values are sampled. This feature is used to reduce the scan time.

The phase-encoding values that are not collected are approximated from those that are. The MR data have special properties that make this possible. That is, the positive and negative phase-encoding values contain similar information. So if the negative 64th phase-encoding value is collected, the positive 64th phase-encoding value can be estimated. This is true for all the phase-encoding values. So, an image can be constructed if as few as half the phase-encoding values are collected - and the other half are derived using information from the half that are collected. Acquired data for full and $\frac{1}{2} N_y$ Fourier acquisitions are shown in Figs. 2a and 2b, respectively.

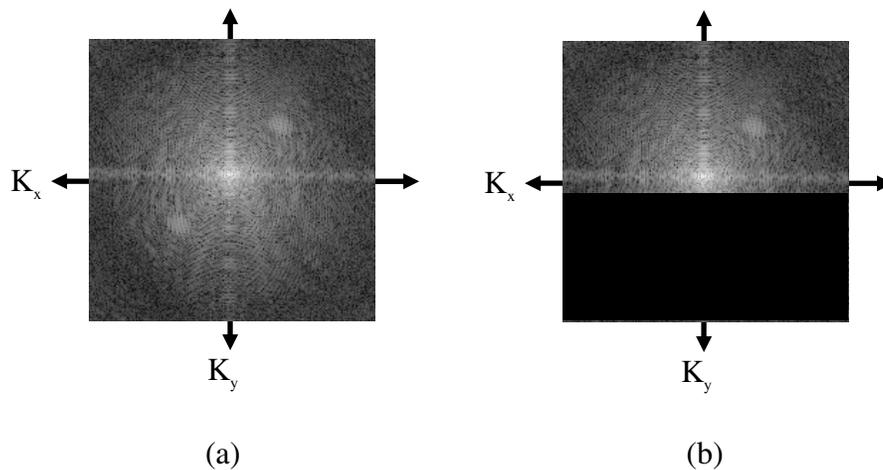


Figure 2: Acquired data for (a) full and (b) $\frac{1}{2} N_y$ Fourier acquisitions. The black region represents lines of Fourier data that are not acquired but are calculated during reconstruction. For $\frac{1}{2} N_y$ Fourier acquisitions, slightly more than half the data must be acquired to aid in calculating the missing data.

The SNR of partial N_y Fourier images is less than that of full N_y Fourier images because only a fraction of the data is acquired for the former. Reviewing the SNR equation, it can be determined that the SNR for $1/2 N_y$ Fourier acquisitions is $\sqrt{1/2}$ times what it is for full N_y Fourier acquisitions. This is because the *voxel volume* does not change (because the missing data are calculated to restore the pixel size), but the *acquisition time* is reduced to $1/2$ because only half of the phase encoding values (N_y) are acquired. Thus, partial N_y Fourier acquisition should only be used when the benefit of the reduced scan time outweighs the penalty of reduced SNR. Other fractions of N_y may be acquired as well (between $1/2$ and full), and the SNR scales according to the square root of the fraction used.

It should be noted that it is also possible to substitute zeroes for all the data that is not acquired. The reconstruction will produce an image with the same number of pixels as a full N_y Fourier acquisition, but the SNR properties are different than described above.

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Traditionally, in MR imaging, the frequency-encoding data are collected symmetrically about the echo. That is, the same number of data samples is collected before the center of the echo as is collected after the center of the echo. It is however possible to produce an image using data that are acquired asymmetrically about the echo. This is what is done with partial Fourier acquisition in the frequency encoding dimension. With partial Fourier acquisition in the frequency encoding dimension, fewer data samples are acquired before the center of the echo than after. These acquisitions are often referred to as asymmetric-, fractional-, or partial-echo acquisitions.

A partial echo is achieved by reducing the duration of the dephasing lobe of the readout (frequency-encoding) gradient. This reduces the amount by which the spins are dephased, so when the refocusing lobe is turned on, the echo is formed sooner than it would have been had the full dephasing gradient been applied. The result is that not all of the negative frequency-encoding values are generated or sampled. After the echo is formed, the readout gradient remains on until all of the positive frequency-encoding values are acquired. The negative frequency-encoding values that are not acquired are calculated in the same way that data points are calculated when partial N_y Fourier acquisition is used. Acquired data for full and $1/2$ echo acquisitions are shown in Figs. 3a and 3b, respectively.

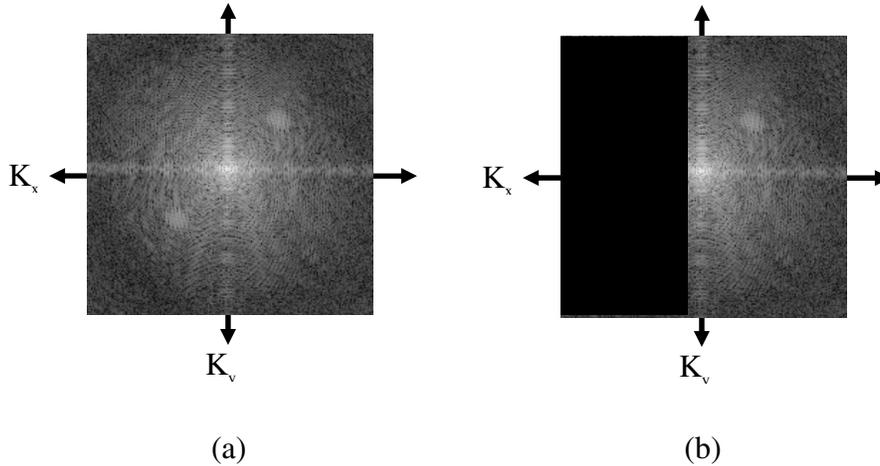


Figure 3: Acquired data for (a) full and (b) $\frac{1}{2}$ echo acquisitions. The black region represents frequency-encoded data that is not acquired but is calculated during reconstruction. For $\frac{1}{2}$ echo acquisitions, slightly more than half the data must be acquired to aid in calculating the missing data.

The advantage of partial echo acquisitions is that they permit shorter echo times to be achieved. This results for two reasons. First, the time from the beginning of data acquisition to the center of the echo decreased, and second, the duration of the readout dephasing lobe is decreased. An additional benefit in gradient echo scans is that the TR is also reduced.

The SNR of partial echo images is less than that of full echo images because only a fraction of the data is acquired. Reviewing the SNR equation, it can be determined that the SNR for $\frac{1}{2}$ echo acquisitions is $\sqrt{\frac{1}{2}}$ times what it is for full echo acquisitions. This is because the *voxel volume* does not change (because the missing data are calculated to restore the pixel size), but the *acquisition time* is reduced to $\frac{1}{2}$ because only half of the frequency-encoding values (N_x) are acquired. However, the reduced signal loss resulting from the reduced echo time, especially in MR angiography methods, often more than compensates for the SNR loss predicted by the above equation alone. Other echo fractions may be used as well (between $\frac{1}{2}$ and full), and the SNR scales according to the square root of the fraction used.

It is worth noting that, care should be taken when using partial Fourier acquisition in both the phase- and frequency-encoding dimensions since this leads to a situation in which not all of the missing data can be estimated. This is because at least half of the data set must be acquired in order to calculate the other half. Using partial Fourier in both the phase- and frequency-encoding dimensions simultaneously results in acquisition of less than half of a data set, preventing determination of some of the missing data. Thus, the image must be reconstructed with a partial data set, which leads to reduced spatial resolution in certain dimensions.

Phase Wrapping

If the object being imaged extends beyond the imaging field-of-view in the phase encoding direction (FOV_y), the signal from outside the FOV_y will appear in the image, but it will be wrapped around to the opposite side, as shown in Fig. 4a. The signal from outside the FOV in the phase-encoding direction can be prevented from appearing in the image, as shown in Fig. 4b, by selecting the no phase wrap (NPW) option.

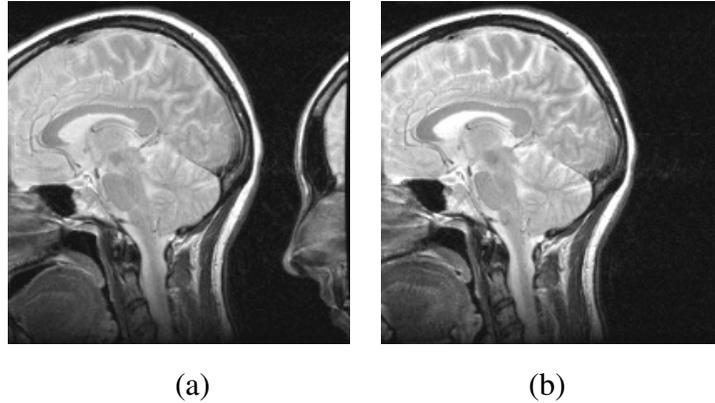


Figure 4: Images acquired (a) without and (b) with no phase wrap (NPW). The wrap around, or aliasing, of signal from outside the FOV_y to the opposite side of the image is eliminated in (b) when NPW is used.

When no phase-wrap is selected, the field-of-view in the phase encoding dimension (FOV_y) is doubled in an attempt to prevent any signal from being outside the FOV_y in this dimension. In order to double the FOV_y in the phase-encoding dimension and at the same time maintain the prescribed spatial resolution (pixel size), the number of phase-encoding values must also be doubled. In order to maintain the prescribed scan time, despite doubling the number of phase-encoding values, $\frac{1}{2}$ the number of averages are acquired. So, when NPW is selected, FOV_y is doubled, N_y is doubled, and N_{ave} is cut in half.

When the image is reconstructed, as long as the object is confined to a region defined by twice the prescribed FOV_y , the object will be accurately represented in the doubled- FOV_y image. Before the image is displayed, the portion outside the prescribed FOV_y is discarded, and only the central portion, free of any aliased signal, will be displayed. Thus, an image of the proper FOV_y and spatial resolution (pixel size) is displayed without aliasing, and the data for the image is collected in the prescribed scan time.

A remarkable feature of NPW, is that, despite the changes in the imaging parameters, the SNR of the images remains unaffected. This is because the *voxel volume* remains unchanged (FOV_y and N_y both double, resulting in no net change to the pixel dimension), and the *acquisition time* remains unchanged (N_y doubles, and N_{ave} is cut in

half). Some vendors allow the FOV to be increased by factors other than two. Appropriate changes can be made to N_y and N_{ave} to maintain the same SNR that would be achieved without the use of NPW.

Chemical Shift

Fat has a short T1 and so it appears bright in most MR images. In some instances, the high signal intensity of the fat makes interpretation of some of the lower signal intensity structures difficult. Also, many times imaging parameters are chosen to make the signal from objects of interest very bright. In these instances, the bright signal from fat can be difficult to discern from the objects of interest. In addition, the signal from fat is misplaced in images due to its chemical shift. This misplacement can cause the signal from fat to overlap with signal from other structures, making it difficult to observe the other structures. For all of these reasons, in MRI, it is often desirable to eliminate the signal from fat.

The signal from fat originates from the nuclei of hydrogen atoms on the fat molecule. In addition to hydrogen atoms, the fat molecule contains carbon atoms. The carbon atoms shield the hydrogen nuclei from the external magnetic field. So, according to the Larmor equation, the signal from fat has a lower resonance frequency than the signal from water. Thus, even when there are no magnetic field gradients turned on, the magnetization from fat and water precess at different frequencies. At 1.5 Tesla, the resonance frequency of the signal from fat is approximately 220 Hz lower than the resonance frequency of the signal from water.

In one fat saturation method, a frequency-selective rf pulse is applied to tip only the magnetization from fat into the transverse plane. Immediately following termination of the rf pulse, a large magnetic field gradient is applied to dephase the transverse magnetization from fat. At completion of the fat saturation pulse (the rf and the dephasing gradient), there is no longitudinal magnetization and no net transverse magnetization from fat. Immediately following the fat saturation pulse, the imaging excitation rf pulse is applied. This tips the longitudinal magnetization from only water into the transverse plane (there is no magnetization from fat at this point in time). The transverse magnetization gives rise to the signal that is used to produce the MR image, and it is void of any magnetization from fat.

The fat is selectively tipped into the transverse plane using an rf pulse that is made up of frequencies that only affect the hydrogen nuclei on the fat molecules. Thus, at 1.5 Tesla, the fat-selective rf pulse has a central frequency that is 220 Hz less than the central frequency of the rf pulse that is used for imaging. There is no need to apply a selection gradient during application of the fat-selective rf pulse because the fat naturally precesses at a different frequency than water.

Images of a breast acquired without and with fat saturation are shown in Figs. 5a and 5b, respectively. The signal from the fat is reduced substantially when fat saturation is used as shown in Fig. 5b.

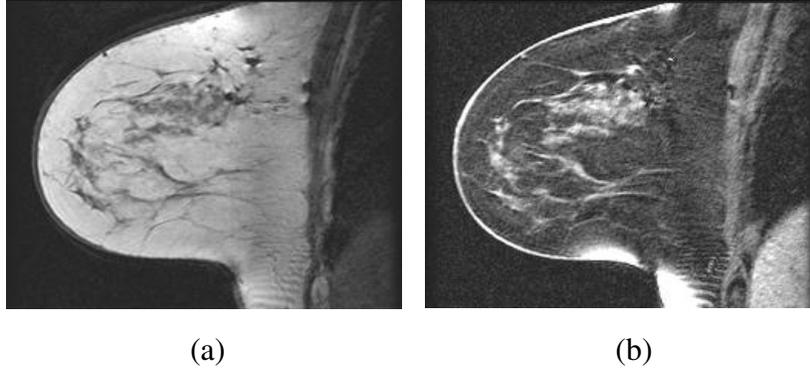


Figure 5: Images of a breast acquired (a) without and (b) with fat saturation. Note that in (b) some signal from fat remains at the inferior aspect of the breast and in the chest wall, because for some reason there is a magnetic field inhomogeneity, causing the fat in this region to have precessional frequencies outside the range affected by the fat saturation pulse.

When fat saturation is performed, the fat saturation pulse (rf and dephasing gradient) is applied prior to each imaging rf excitation pulse. The long duration of the fat saturation pulse causes the minimum TR to increase for sequentially-acquired gradient-echo scans, and the maximum number of slices that can be imaged in a given TR to decrease for spin-echo scans. Other fat saturation methods exist, but are not discussed here.

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Spatial saturation is an option that permits elimination of unwanted signal from a particular region within the coil; either inside or outside the FOV. It is used for example to eliminate signal from pulsatile blood or signal in the portion of the abdominal wall that moves during respiration so that the ghosts that would otherwise originate from these objects do not mimic or obscure visualization of pathology.

The spatial saturation pulse consists of an rf pulse applied simultaneously with a slab selection gradient, followed immediately by a large dephasing gradient. The spatial saturation pulse works as follows. The rf pulse, applied simultaneously with the slab-selection gradient, tips the magnetization in a specified region into the transverse plane. Immediately following this, a large gradient is applied to dephase the transverse magnetization in the specified region. Thus, at the conclusion of the spatial saturation pulse, no longitudinal or net transverse magnetization remains in the specified region (the saturated band). Immediately following the spatial saturation pulse, the imaging excitation rf pulse is applied and the signal is sampled. Because no magnetization is available from the region affected by the saturation pulse, this region gives rise to no signal when the imaging excitation rf pulse is applied. If the region affected by the spatial saturation pulse is within the FOV, this region will show up as a signal void in the image. If the spatial saturation pulse is applied outside the FOV or outside the slice, any blood that experiences the saturation pulse and then moves into the region being imaged before

the imaging excitation rf pulse is applied will give rise to no signal. The saturation pulse is applied before every excitation pulse to make sure that there is no signal from the saturation band at the time the imaging rf excitation pulse is applied. The user can specify the thickness and the location of the saturation band. The effect of placing the saturation band over the abdominal wall to suppress ghosts induced by respiratory motion is demonstrated in sagittal T1 and T2 images of the spine shown in Figs. 6a and 6b, respectively.

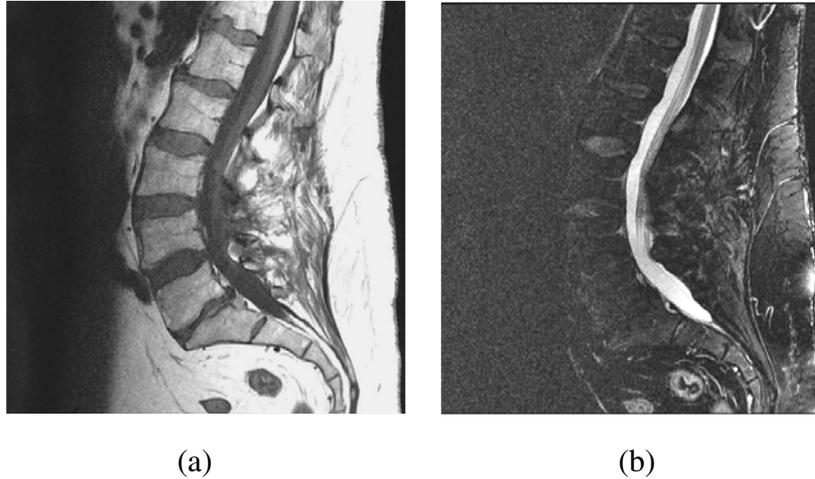


Figure 6: Sagittal (a) T1 and (b) T2 images of the spine acquired with a spatial saturation band applied at the anterior aspect of the FOV to eliminate signal, and therefore ghosts, that might have otherwise originated from the abdominal wall due to respiratory motion.

When spatial saturation is performed, the spatial saturation pulse (rf, slab selection gradient, and dephasing gradient) is applied prior to each imaging rf excitation pulse. The added time necessary to apply the spatial saturation pulse causes the minimum TR to increase for sequentially-acquired gradient-echo scans, and the maximum number of slices that can be imaged in a given TR to decrease for spin-echo scans.

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There are a great number of imaging options available, far more than are described here. There exist imaging options to reduce or eliminate artifacts, decrease scan time, and improve image contrast. Knowing how the imaging options work, how they affect the acquisition, and how they influence image quality allows them to be used effectively.