Fast Spin Echo Imaging
Frank R Korosec, PhD
University of Wisconsin – Madison

In conventional spin echo (SE) imaging, the longitudinal magnetization (that is generated by the presence of the main magnetic field, B_0) is tipped into the transverse plane using a radiofrequency (RF) pulse that tips the magnetization 90 degrees. Once in the transverse plane, the magnetization rotates, or precesses, around the axis of the main magnetic field. As the transverse magnetization precesses, it generates a time-varying signal in the receiver coil. While the transverse magnetization is precessing, it is diminishing. Two phenomena that cause diminution of the transverse magnetization, and the signal generated by it, are interactions with nuclear neighbors (T2 relaxation) and magnetic field inhomogeneities (T2* relaxation). T2 relaxation is desirable in that it permits signal differences from tissues with different T2 values. T2* can be detrimental as it causes loss of signal in regions affected by magnetic field inhomogeneities. T2* effects can be ameliorated by applying an RF pulse that flips the transverse magnetization 180 degrees. When this is done, the magnetization that is lost (dephased) prior to the application of the 180 degree RF pulse is recovered (rephrased) after the application of the 180 degree pulse. When the maximum magnetization is restored at the echo time (TE), the receiver is turned on and the signal is sampled. During the application of the 90 and 180 degree RF pulses, a magnetic field gradient is applied so that the RF pulses tip only magnetization in a given slice. This magnetic field gradient is referred to as the slice-selection gradient.

In order to produce images, the signal generated by the transverse magnetization must be encoded to contain information regarding the location from where it originated. In one dimension, the magnetization from different locations is made to precess at different frequencies so that signals from different spatial locations oscillate at different rates. This is achieved by applying a magnetic field gradient (a magnetic field that changes linearly with position) during the detection of the signal so that magnetization at different locations experiences different magnetic fields and therefore precesses at different frequencies. In the second dimension, magnetization from different locations is made to precess by different amounts so that when the receiver is turned on and the signals are sampled, signals from different locations have a different starting phase since the magnetization from different locations is rotated by different amounts by the time the receiver is turned on. The magnetic field gradient used to encode position based on the oscillatory frequency of the signals is referred to as the frequency encoding gradient, and the magnetic field used to encode position based on the starting phase of the signals is referred to as the phase-encoding gradient.

The 90 degree tipping RF pulse combined with the 180 refocussing RF pulse make up what is referred to as the spin echo sequence. Adding the slice-selection, the frequency-encoding, and the phase-encoding gradients, makes it a spin echo imaging sequence. To create an image using this method requires sampling many phase-encoded spin echoes. Typically, one phase-encoded echo is acquired per repetition of this sequence. The

repetition time (TR) determines the image acquisition time according to the following expressing:

$$\text{Acquisition Time (SE)} = TR \times PE \times NSA,$$

where TR is the repetition time, PE is the number of phase-encoded spin echoes sampled, and NSA is the number of signal averages (the number of times each phase-encoded spin echo is sampled and averaged to improve image quality).

The repetition time (TR) is determined by the type of image contrast desired. In a T1-weighed image acquisition, it is necessary to use a short TR (600 ms = 0.6 sec), whereas in a T2-weighted scan, it is necessary to use a long TR (3000 ms = 3 sec). Thus, for a scan requiring 256 phase-encoded spin echoes and one signal average, the time required to acquire a T1-weighed image set would be 2 minutes and 34 seconds, and the time required to acquire a T2-weighted image set would be 12 minutes and 48 seconds. If more signal averages are required, or more coverage is desired, the scan times increase accordingly.

With fast spin echo (FSE) imaging, the scan time is reduced by acquiring multiple phase-encoded spin echoes per TR. So, the fast spin echo method begins the same as the conventional spin echo method. In the conventional spin echo method, the transverse magnetization is allowed to dephase after the spin echo signal is sampled. In the next TR, more transverse magnetization is generated and the sequence of RF and gradient pulse is repeated so that another phase-encoded spin echo may be produced and sampled. However, in fast spin echo imaging, after the first phase-encoded spin echo is sampled, another 180 degree refocusing pulse is applied to refocus the transverse magnetization that began to dephase after the first phase-encoded echo was sampled. Just prior to the second spin echo (formed by the second 180 degree RF pulse), the phase encoding from the first phase-encoding gradient is undone (by applying a negative phase-encoding gradient), and a second different-sized phase-encoding gradient is applied. This second phase-encoded spin echo is then sampled by turning on the receiver briefly. This process can be repeated by applying more 180 degree refocusing RF pulses, and more different-sized phase-encoding gradients, to sample multiple phase-encoded spin echoes during a single TR. This permits acquisition of a full image data set in fewer TR intervals. So, the acquisition time required using a fast spin echo imaging sequence is as follows:

$$\text{Acquisition Time (FSE)} = TR \times PE/ETL \times NSA$$

where TR is the repetition time, PE is the number of phase-encoded spin echoes produced and sampled, ETL is the echo train length (the number of phase-encoded spin echoes produced and sampled during each TR), and NSA is the number of signal averages as described above.

So, a major advantage of fast spin echo imaging is the reduction in scan time it offers. Typical echo train lengths are between 4 and 16, with concomitant reductions in acquisition time. For some applications, all of the phase-encoded spin echoes for a single

image can be acquired in a single TR. With recent advances in MRI technology and innovations in pulse sequence design, echo train lengths can be very long, making 3D fast spin echo acquisition achievable in reasonable scan times.

Obtaining the proper contrast in fast spin echo images is achieved by strategically applying the phase-encoding gradients at the appropriate times during the echo train. Because the 180 degree RF pulses do not refocus the signal loss caused by T2 relaxation (they only refocus signal losses due to T2* relaxation), the signals from all tissues are similar at the first echo in the train, but the signals from different tissues can be very different at the later echoes in the train. For example, tissues with long T2 relaxation times will retain much signal when the late echoes in the train are acquired whereas tissues with short T2 relaxation times will lose much signal by the time the late spin echoes in the train are acquired. So, tissues with different T2 values will give rise to similar signals early in the echo train, and dissimilar signals late in the echo train.

It turns out that the information content in the low order phase-encoded spin echoes (produced using small phase-encoding gradient amplitudes) is largely responsible for the contrast (signal differences from different tissues) in the images, whereas the information content in the high order phase-encoded spin echoes (produced using large phase-encoding gradient amplitudes) is largely responsible for the detail (tissue edges) in the images.

So, if the desire is to acquire T2-weighted images with a fast spin echo imaging method, the phase-encoding gradients are arranged so that the small phase-encoding gradients (used to produce signals responsible for image contrast) are applied late in the echo train. In this way, the low order phase-encoded spin echoes (small phase-encoding gradients) are acquired when the tissues with different T2 values have achieved signal differences. Conversely, the large phase-encoding gradients are applied early in the echo train, so, unfortunately, the detail in the images will not contain much T2 weighting (signals from all tissues are nearly the same early in the echo train, independent of the T2 of the tissues). However, this does not affect the overall contrast of the image too much, and is well tolerated.

Alternatively, if the desire is to reduce signal differences (contrast) caused by differences in T2 values among different tissues (as is the case when producing T1- or proton-density-weighted images), the small phase-encoding gradients are applied early in the echo train so these values are collected before any signal differences develop due to differences in T2 relaxation rates. The large phase-encoding gradients are applied late in the echo train, so, despite the fact that signal differences have developed in tissues with different T2 relaxation rates, these signal differences will only affect the edges of the tissues. So, the overall image contrast is largely influenced by the small phase-encoding gradients that are collected early in the echo train before any T2-related signal differences can manifest. In this way, T1- and density-weighted images can be kept largely free of signal differences caused by T2 relaxation (except at tissue edges).

As previously described, with fast spin echo imaging, as the multiple echoes are being generated, phase encoded, and sampled in the echo train, the signals from all tissues are diminishing due to T2 relaxation. This means that that typically there is more signal from all tissues at the beginning of the echo train, and less signal from all tissues at the end of the echo train. These differences in signal throughout the echo train affect the appearance of the images.

Recall that with T1-weighed fast spin echo imaging the contrast information (small phase-encoding gradients) is collected early in the echo train, whereas the detail information (large phase-encoding gradients) is collected late in the echo train. This means that the signals used to determine the detail information are diminished relative to the signals used to determine the contrast information. This leads to a suppression of edge information in T1-weighed images acquired using fast spin echo imaging methods. For this reason, in order to limit blur, the length of the echo train used for T1-weighted images is limited to about 3 or 4 echoes so that not too much signal diminution is permitted to occur during the echo train.

With T2-weighted fast spin echo imaging, the detail information (large phase-encoding gradients) is collected early in the echo train, whereas the contrast information (small phase-encoding gradients) is collected late in the echo train. This means that the signals used to obtain the detail information are bright (because they are collected early in the echo train before much T2 relaxation has occurred), leading to enhancement of the detail information (edge enhancement) in T2-weighed images acquired using fast spin echo imaging methods.

There are numerous other considerations associated with using fast spin echo imaging methods that will be discussed during this presentation. Additionally, there are numerous variants of the fast spin echo imaging method that may be employed for various applications. For example, for certain applications, all of the data for a single image may be acquired in a single echo train. In another application, in order to enhance T2 contrast and shorten the TR of the imaging sequence, an additional 180 degree pulse and then a negative 90 degree RF pulse may be used to refocus and then tip the remaining transverse magnetization back along the longitudinal axis, so the longitudinal magnetization contains T2-weighting, and it is immediately ready for the next TR (no need to wait for T1 relaxation.) For applications requiring added T1 contrast, an inversion pulse may be applied prior to each fast spin echo readout interval. For applications in which the signal from CSF needs to be eliminated, an inversion pulse may be applied prior to the fast spin echo readout interval, and a TI can be selected so that the fast spin echo readout begins when the CSF is going through its zero crossing point. In cardiac imaging, two IR pulses and appropriate TI times can be used to null the signal from blood, or three IR pulses and appropriate TI times can be used to null signals from both blood and fat. Using a very long TE with a fast spin echo imaging method allows acquisition of MRCP images. And, as mentioned previously, with strong and fast-switching magnetic field gradients, manipulation of the RF tip angles throughout the echo train, and some other innovations, very long echo trains can be used to permit 3D fast spin echo scans to be performed in

reasonable acquisition times. These variants of the fast spin echo method will be briefly described.

Fast spin echo imaging methods are useful for reducing the time required for acquiring image data. By appropriately arranging the phase-encoding values along the echo train, the desired contrast may be obtained. The goal of this presentation is to provide some basic information about the principles of the fast spin echo method and it variants so that users may effectively utilize these methods.