MRI: spatial localization

When the patient is placed in the MRI scanner, the bulk magnetisation of the protons in the patient develops as spins aligned with the main magnetic field (Bo). For imaging, this magnetisation needs to be tipped away from Bo, which is done by applying an RF pulse (1). The tipped magnetisation relaxes, emitting RF signal at the larmor frequency. In a homogeneous magnetic field, all the protons in the body would spin at the larmor frequency and the signal received will contain a complex mix of frequencies, phases and amplitudes. As the entire body emits this signal, it must be isolated in some manner so that its origin in three dimensions can be determined. To obtain the spatial information, we manipulate the magnetic field in the X, Y and Z directions, which is done by the application of magnetic field gradients. Magnetic field gradients are generated by gradient coils of which there are three pairs (X, Y and Z). The gradient coils are switched at different times in different directions to perform spatial localization.

Slice selection is the first step of spatial localisation. This is performed by the Z-gradient, also referred to as the slice select gradient. To excite the slice selectively, the magnetic field along the Z-direction is made non-uniform. The protons in the head will spin faster than at iso-centre and the protons at iso-centre will move faster than the feet. Choosing a particular frequency bandwidth, it is now possible to excite exactly one corresponding slice without influencing the rest of the body. For example at 3T, protons in the feet would spin at 123.3 MHz, the ones at iso-centre would be spinning at 123.4 MHz and the ones in the head at 123.5 MHz. If we apply an RF pulse at 123.5 MHz, only the protons in the thin slice in the head will react because they are the only ones spinning at that frequency. This is slice selection. The amplitude of the slice select gradient determines both the slice thickness and slice position regardless of orientation. Lower gradient strength will lead to thicker slices while stronger gradients are needed to generate thinner slices (Fig. 1). The slice select gradient is applied during the RF pulse.

After selecting a slice using the slice select gradient, we use the phase-encoding and frequency-encoding gradients to get information about individual pixels in that slice.

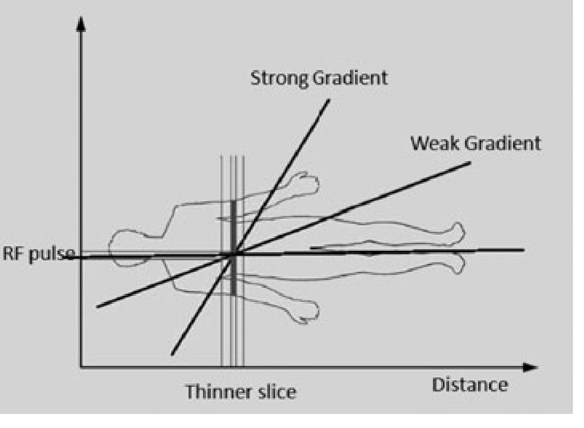

Fig. 1. The strength of slice select gradient determines the slice thickness (figure courtesy of ‘How does MRI work’ by Weishaupt).

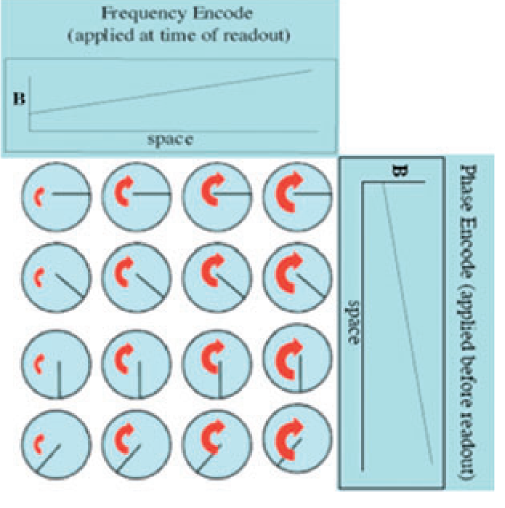

Fig. 2. Frequency and phase-encoding steps.

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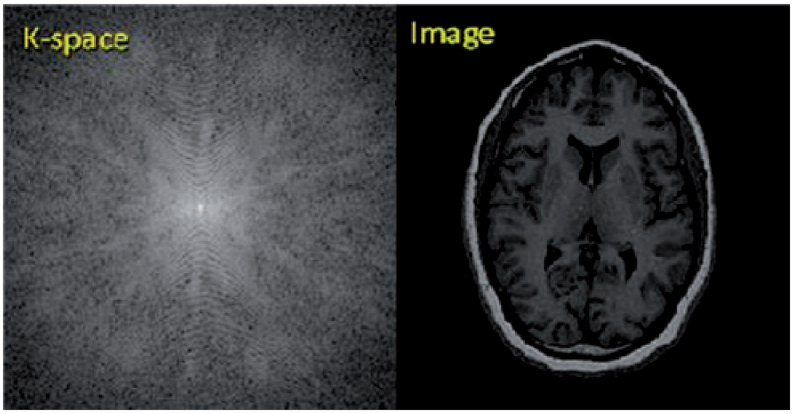

Fig. 3. The MR data shown in the k-space and image space.

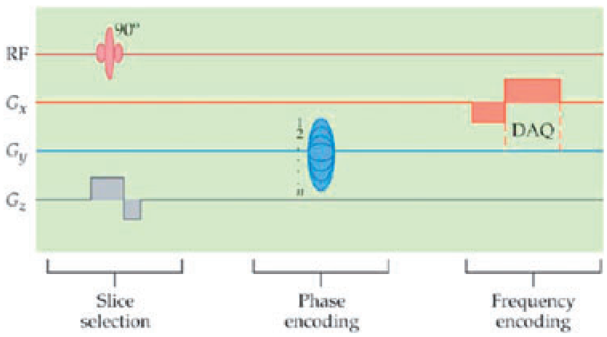

Fig. 4. A basic pulse sequence with X, Y and Z gradients.

To achieve phase encoding, a gradient in the Y-direction (from top to bottom) will be switched on after the RF pulse ((2)). This causes the larmor frequency at the top to be slightly higher than the bottom of the slice. As the spins are already excited (by the RF pulse) and are precessing in the XY plane, those at the top will spin somewhat faster than the ones further down. This results in a phase shift between the spins. The Y-gradient uses phase encoding to divide the slice in to horizontal rows.

Till now, we could determine two things: the signal comes from the head (slice encoding) and the signal contains a number of RF waves, which have the same frequency but different phases. It is possible to tell whether the signal comes from anterior or from posterior (phase encoding). All that remains is to determine whether the signal comes from the left, centre or the right side of the head. This is be done by the frequency-encoding step.

Frequency encoding is done by the X-gradient, also called the readout gradient. The X-gradient coil looks exactly like the Y-gradient but is rotated 90°. At the end of the phase-encoding step, all spins return to precessing at larmor frequency associated with the main magnetic field. The purpose of X-gradient is to change, very slightly, the magnetic field in the imaging volume to create a gradient in the X-axis (left to right). The X-gradient will cause the protons in the vertical columns to experience slightly different magnetic fields and thus precess at different frequencies. The protons on the left-hand side spin with a lower frequency than the ones on the right. They will accumulate an additional phase shift because of a different frequency. Using this phase shift information in addition to the acquired phase difference by the phase-encoding gradient will determine the spatial position of the MR signal within the selected slice (Fig. 2).

The RF signal received during the frequency encoding is then used to fill the k-space. k-Space can be thought of as a digitised version of the data space. The centre of the k-space will always contain the phase-encoding steps with the weakest gradient and thus more signal. The periphery of k-space will contain those phase-encoding steps with the largest gradients and thus with the least signal (3). Each data point in k-space consists of the summation of MR signal from all voxels in image space under corresponding gradient fields. The inverse Fourier transform of the k-space will generate an image. Figure 3 shows the MR data in the k-space and image space. The sequence of events that occur to create an image is called a pulse sequence. It is a timing diagram that shows the RF pulses, gradients and echoes. Figure 4 shows a basic pulse sequence.

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