Relaxation Times, Gradient Echoes and Spin Echoes

5

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Abstract

There are two distinct relaxation processes that relate to the two components of the net magnetisation. The first longitudinal relaxation process, commonly referred to as T1 relaxation is responsible for the re-growth of the *z* -component along the longitudinal (*z*) axis to its original value at equilibrium. The second transverse relaxation process is responsible for the decay of the *xy* component as it rotates about the *z* -axis and hence the observed decay of the MR signal. There are two types of transverse relaxation, known as T2 relaxation and T2* relaxation. T2 relaxation is caused by a loss of coherence of the rotating proton spins due to the interaction neighbouring magnetic moments (spin-spin interactions). This is an exponential process with a time constant, T2. Signal decay is also caused by a loss of coherence between rotating spins due to non-uniformities in the magnetic field This together with T2 relaxation leads to a more rapid decay known as T2* relaxation with a time constant, T2*. For MR imaging the MR signal generated and measured in the form of an echo. Gradient echoes are generated by switching the direction of a magnetic field gradient to first de-phase then re-phase the MR signal. Spin echoes are generated by applying a 180° RF refocusing pulse to reverse the decay caused by magnetic field inhomogeneities. The time from the RF pulse to the maximum amplitude of the echo is known as the echo time, TE.

Keywords

Relaxation • Longitudinal relaxation • Transverse relaxation • T1 relaxation • Saturation recovery • Molecular tumbling rate • Macromolecular content • T2 relaxation • T2* relaxation • Spin-spin relaxation • Magnetic field inhomogeneities • De-phasing • MR echoes • Gradient echoes • Spin echoes • Echo time • TE

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Relaxation: What Happens After the RF Excitation Pulse?

Immediately after the RF pulse the spin system starts to return back to its original state, i.e., equilibrium. This process is known as *relaxation*. In fact there are two distinct relaxation processes that relate to the two components of the net magnetisation, the longitudinal (*z*) and transverse (*xy*) components. The first *longitudinal relaxation* process, commonly referred to as *T1 relaxation* is responsible for the re-growth of the *z* component along the longitudinal (*z*) axis to its original value at equilibrium. This is explained more fully in the next section. The second *transverse relaxation* process is responsible for the decay of the *xy* component as it rotates about the *z*-axis and hence the observed decay of the MR signal. There are two types of transverse relaxation, known as *T2 relaxation* and *T2* relaxation* and these are explained more fully later in this chapter. Longitudinal and transverse relaxation both occur at the same time, however, transverse relaxation is typically a much faster process in human tissue. i.e., the signal decays away long before the spin system returns to equilibrium.

What is T1 Relaxation?

T1 relaxation describes the recovery of the *z*-component (M_z) of the magnetisation following an RF pulse as the population of protons returns to its equilibrium state. In the previous example of a 90° pulse (*saturation pulse*), the *z*-magnetisation is *saturated* (reduced to zero) immediately after the pulse, but then returns along the *z*-axis towards its equilibrium value initially rapidly, slowing down as it approaches its equilibrium value. The return of the M_z component along the *z* -axis is an exponential process with a time constant *T1* (Fig. 5.1). The shorter the T1 time constant is, the faster the relaxation process and the return to equilibrium. Recovery of the z-magnetisation after a 90° RF pulse is sometimes referred to as *saturation recovery*.

What's the Significance of the T1 Value?

T1 relaxation involves the release of energy from the proton spin population as it returns to its equilibrium state. The rate of relaxation is related to rate at which energy is released to the surrounding molecular structure. This in turn is related to the size of the molecule that contains the hydrogen nuclei and in particular the rate of molecular motion, known as the tumbling rate of the particular molecule. As molecules tumble or rotate they give rise to a fluctuating magnetic field which is experienced by protons in adjacent molecules. When this fluctuating magnetic field is close to the Larmor frequency, energy

exchange is encouraged. For example, lipid molecules are of a size that gives rise to a tumbling rate which is close to the Larmor frequency and therefore extremely favourable for energy exchange. Fat therefore has one of the fastest relaxation rates of all body tissues and therefore the shortest T1 relaxation time. Larger molecules have much slower tumbling rates that are unfavourable for energy exchange, giving rise to long relaxation times. For free water, its small molecules have much faster molecular tumbling rates which are also unfavourable for energy exchange and therefore it has a long T1 relaxation time. The tumbling rates of water molecules that are adjacent to large macromolecules can however be slowed down towards the Larmor frequency shortening the T1 value. Water-based tissues with a high macromolecular content (e.g., muscle) tend to have shorter T1 values. Conversely, when the water content is increased, for example by an inflammatory process, the T1 value also increases.

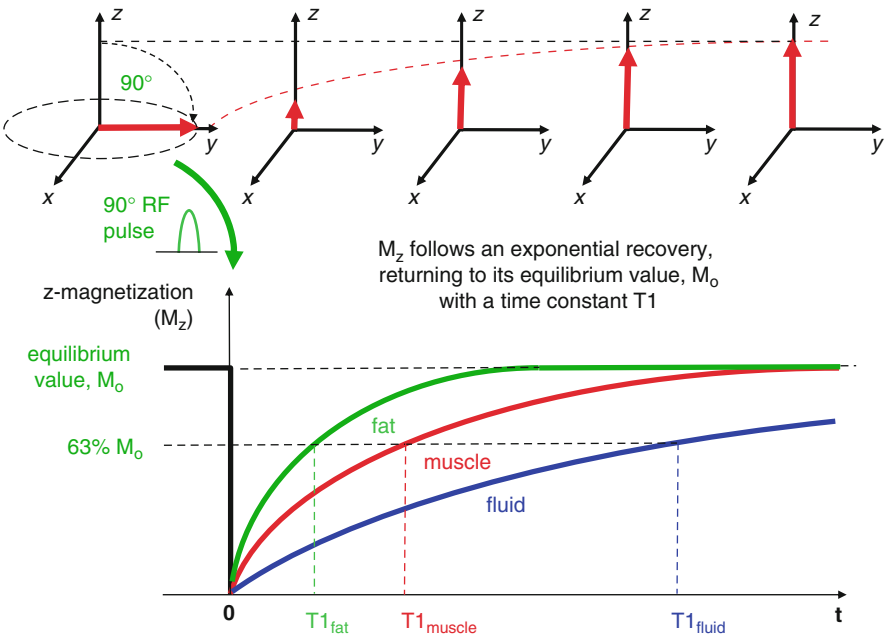

Fig. 5.1 Following a 90 RF pulse, the z component of the Net Magnetisation, M_z is reduced to zero, but then recovers gradually back to its equilibrium value if no further pulses are applied. The recovery of M_z is an exponential process with a time constant T1. This is the time at which the magnetization has recovered to 63 % of its original value. Different tissues have different T1 values. Fat has the shortest T1 value of all tissues (fat recovers the fastest) while fluid has the longest T1 value (fluid recovers the slowest)

Transverse Relaxation and MR Signal Decay

In Chap. 4 we saw that the decay of the MR signal (or free induction decay) is due to a loss of coherence as the magnetic moments (spins) of proton population move out of phase. The causes of this loss of coherence are twofold:

(i) The presence of interactions between neighbouring protons.
(ii) Local variations (inhomogeneities) in the applied magnetic field.

The transverse relaxation caused by (i) alone is known as *T2 relaxation*. The transverse relaxation actually observed in an FID is the combination of (i) and (ii) and is known as *T2* relaxation*.

What is T2 Relaxation?

The rate of precession for an individual proton depends on the applied magnetic field. It is however possible for the magnetic moment of one proton to slightly modify the magnetic field of a neighbouring proton (Fig. 5.2). As the protons are constituents of atoms within molecules, they are moving rapidly and randomly and so such effects are transient and random. The net effect is for the Larmor frequency of the individual protons to fluctuate in a random fashion, leading to a loss of coherence across the population of protons. i.e., the spins gradually acquire different phase angles, pointing in different directions to one another and are said to move out of phase with one another (this is often referred to as *de-phasing*).

The resultant decay of the transverse component of the magnetisation (M_{xy}) has an exponential form with a time constant, T2, hence this contribution to transverse relaxation is known as *T2 relaxation*. As it is caused by interactions between neighbouring proton spins it is also known as *spin-spin relaxation*. Due to the random nature of the spin-spin interactions, T2 relaxation is irreversible.

What's the Significance of T2 Value?
T2 relaxation is related to the amount of spin-spin interaction that take place. Free water are small molecules that are relatively far apart and moving rapidly and therefore spin-spin interactions are less frequent and T2 relaxation is slow (leading to long T2 relaxation times). Water molecules bound to large molecules are slowed down and more likely in interact, leading to faster T2 relaxation and shorter T2 relaxation times. Water- based tissues with a high macromolecular content (e.g., muscle) tend to have shorter T2 values. Conversely, when the water content is increased, for example by an inflammatory process, the T2 value also increases.

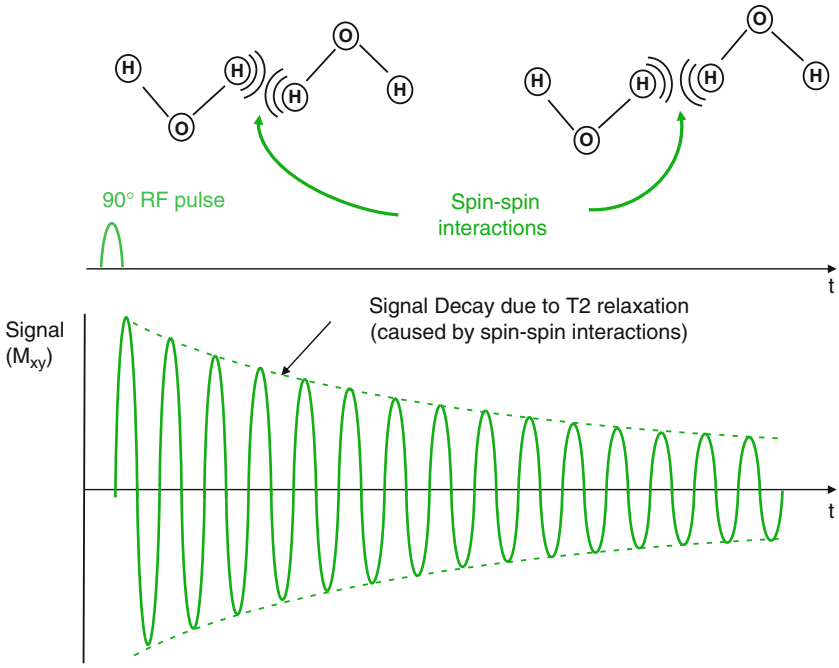

Fig. 5.2 The FID signal decays due to spin-spin interactions. This is where the magnetic moment of one proton moves transiently adjacent to another proton, slightly modifying the local magnetic field of its neighbour and therefore causing the Larmor frequency to briefly alter. This in turn leads to a loss of phase coherence (de-phasing). Due to the random nature of molecular motion, this process is irreversible

What is T2* Relaxation?

The second cause for the loss of coherence (de-phasing) relates to non-uniformities in the applied magnetic field, B_o. If this field varies with position, then so does the Larmor frequency (Fig. 5.3). Protons at different spatial locations will therefore rotate at different rates, causing further de-phasing and the signal to decay more rapidly. In this case, as the cause of the variation in Larmor frequency is fixed, the resultant de-phasing is potentially reversible.

The combined effect of T2 relaxation and the effect of magnetic field non-uniformities is referred to as *T2* relaxation* and this determines the actual rate of decay observed when measuring an FID signal.

MR Echoes

Whilst the FID can be detected as a MR signal, for MR imaging it is more common to generate and measure the MR signal in the form of an echo. This is because the magnetic field gradients that are used to localise and encode the MR signals in space

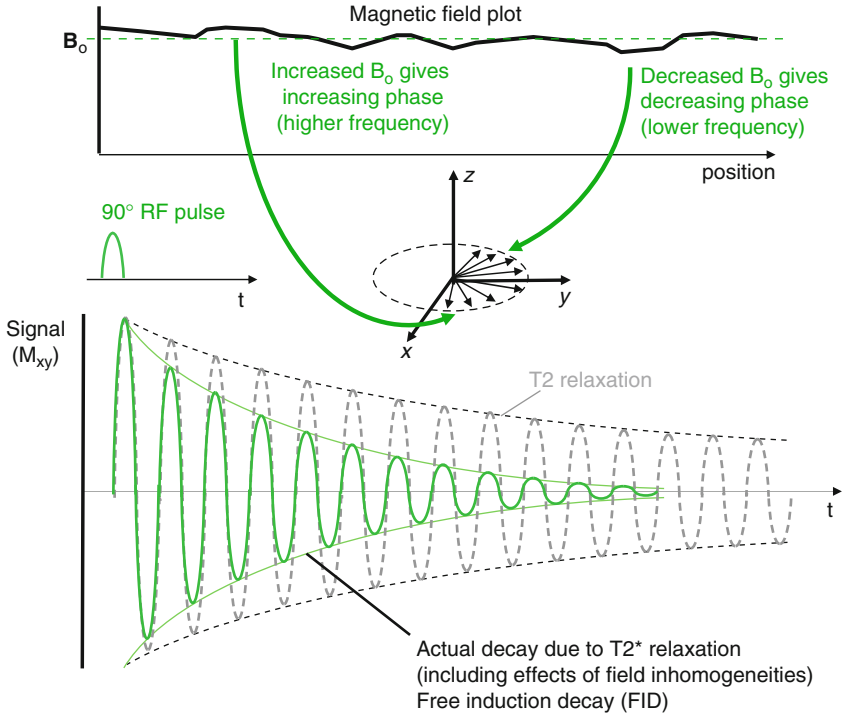

Fig. 5.3 A further cause of loss of phase coherence is that the applied magnetic field, B_o is not uniform. While the nominal magnetic field value is assumed to be uniform (shown by the dotted green line) the actual field value varies. Where the magnetic field is decreased, the Larmor frequency is decreased and the relative phase of the proton magnetic moments decreases over time relative to the proton magnetic moments at the nominal Larmor frequency. Where the field is increased in value, this results in a increasing phase. This causes additional de-phasing and accounts for the more rapid decay of the FID signal, known as T2* decay

(as we shall see in Chap. 6) cause additional de-phasing which disrupts the FID. The two most common types of echo used for MR imaging are *gradient echoes* and *spin echoes*. The following sections describe how these echoes are generated.

Gradient Echoes

Magnetic field gradients are used to produce a change in field strength and hence a corresponding change in Larmor frequency along a particular direction. When a magnetic field gradient is switched on it causes proton spins to lose coherence or de-phase rapidly along the direction of the gradient as they precess at different frequencies (Fig. 5.4). This de-phasing causes the amplitude of the FID signal to rapidly drop to zero.

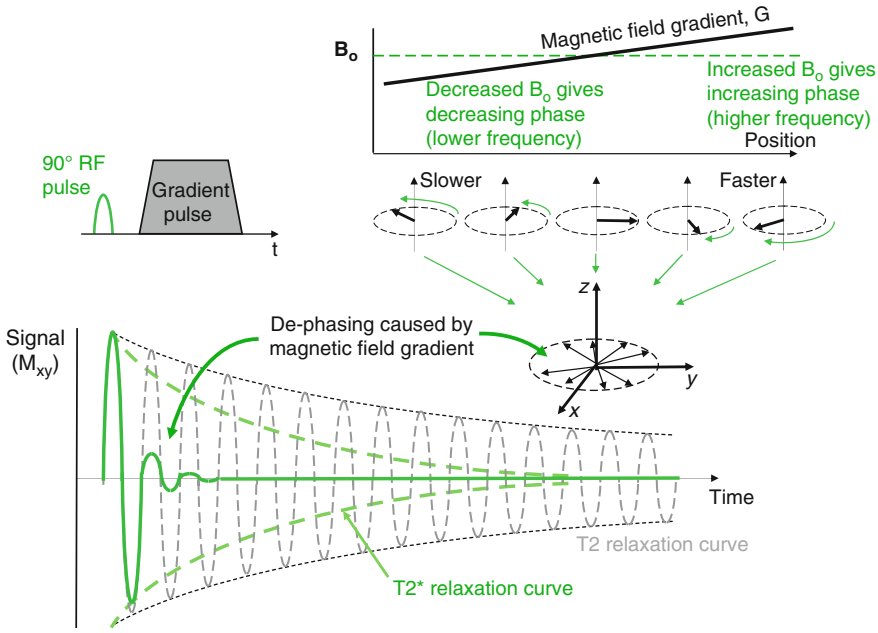

Fig. 5.4 This shows how the application of a magnetic field gradient causes loss of detectable MR signal. The range of frequencies along the direction of the gradient causes the spins to rapidly de-phase, resulting in a rapid reduction of the transverse magnetisation

The amount of de-phasing caused by one magnetic field gradient can however be reversed by applying a second magnetic field gradient along the same direction with equal amplitude but with the opposite slope (Fig. 5.5). If the second gradient is applied for the same amount of time as the first gradient, the de-phasing caused by the first gradient is cancelled and the FID re-appears. It reaches a maximum amplitude at the point at which the spins de-phased by the first gradient have moved back into phase, or 're-phased' (ignoring any effects of T2* relaxation).

If the second gradient then continues to be applied, the FID signal de-phases and disappears once more (Fig. 5.6). The signal that reappears (re-phases) through the switching of the gradient direction is known as a *gradient echo*.

The time from the point at which the transverse magnetisation (the FID) is generated by the RF pulse, to the point at which the gradient echo reaches its maximum amplitude is known as the *echo time* (abbreviated *TE*). If the echo time is chosen to be longer, more natural T2* de-phasing occurs and the maximum echo amplitude becomes smaller. In practice, the TE is set by the MR system operator (in milliseconds) as it determines, amongst other things, the influence of T2* on the image contrast.

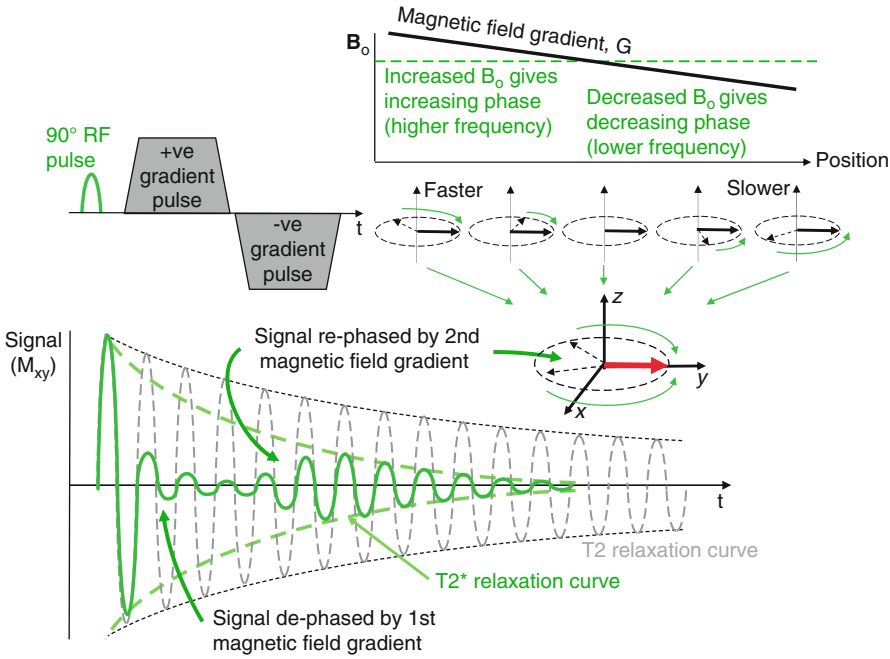

Fig. 5.5 This shows how the application of a second magnetic field gradient reverses the de-phasing caused by the first gradient pulse, resulting in recovery of the FID signal

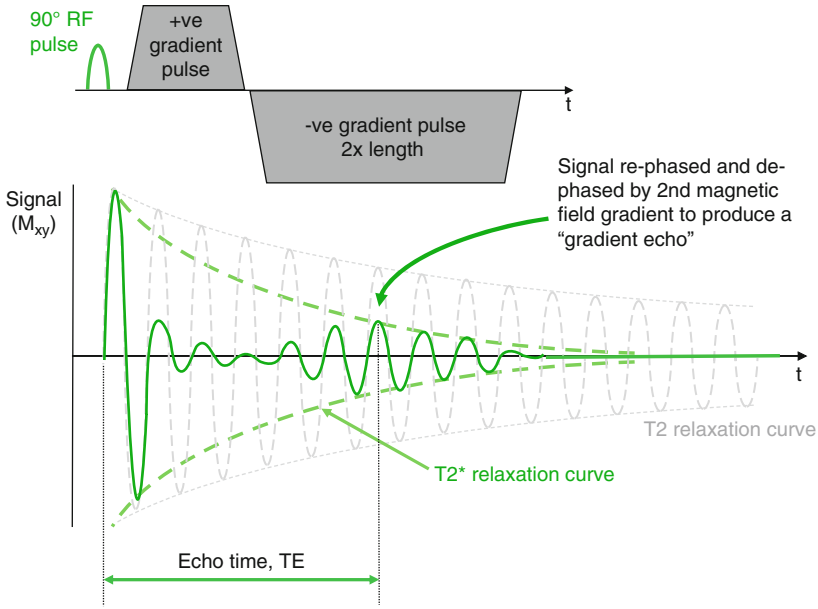

Fig. 5.6 Extension of the time duration of the second gradient to twice that of the first gradient causes the FID to first re-phase and then de-phase. The resultant transient signal is known as a gradient echo. The maximum amplitude of the echo depends on the T2* relaxation rate and the echo time TE

Spin Echoes

Earlier in this chapter we learned that while the de-phasing caused by T2 relaxation was a random, irreversible process, the additional de-phasing caused by the presence of magnetic field non-uniformities was potentially reversible. At a certain time after the initial generation of the FID signal, a proportion of the relative phase change for each proton spin will be related to the local value of the applied magnetic field. The application of a 180° refocusing pulse rotates the spins through 180°, effectively changing the sign of the relative phase change within the *xy* plane (Fig. 5.7). Where the previous relative phase change was positive due to a locally increased field, the 180° pulse causes it to become negative and visa versa. As the local field variations remain fixed, the spins still continue to have the same Larmor frequency, so a spin in an increased field continues to gain in phase, while a spin in a decrease field continues to lose phase. Because the sign of their phase shifts has been swapped halfway through by the 180° refocusing pulse, the spins all come back into phase causing the FID to increase in amplitude, reaching a maximum at the echo time, TE. For the spin de-phasing caused by the field non-uniformities to be completely reversed at time TE, the 180° pulse must be applied halfway through the echo time at time TE/2. The signal that appears (re-phases) through the application of the 180° RF refocusing pulse is known as a *spin echo* (Fig. 5.8). After reaching a maximum amplitude at time TE, the signal again de-phases due to the T2* relaxation process.

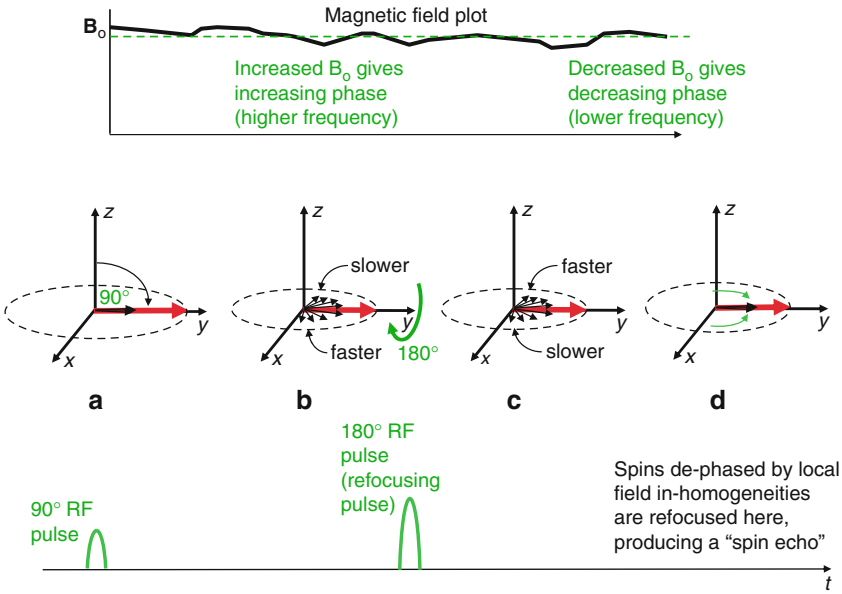

Fig. 5.7 The presence of magnetic field inhomogeneities cause the proton magnetic moments to de-phase (**a**–**b**). The application of a 180° RF pulse causes an instantaneous change in sign of the phase shifts by rotating the spins about the y axis (**b**–**c**). The proton magnetic moments the move back into phase, reversing the de-phasing effect of the magnetic field inhomogeneities (**c**–**d**)

For the purposes of imaging, a magnetic field gradient is also applied during both the de-phasing period and during the measurement of the spin echo (Fig. 5.9). In general, because of the 180° refocusing pulse, the amplitude of the spin echo signal

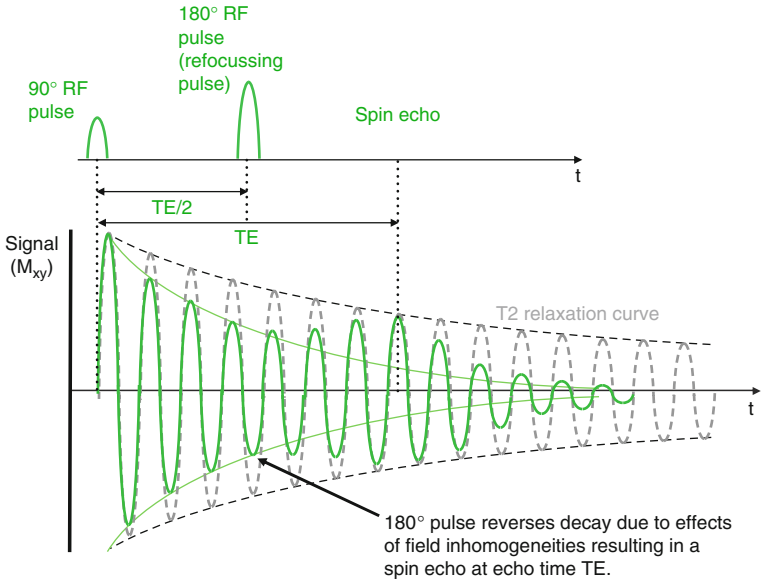

Fig. 5.8 The MR signal that is refocused by the 180° pulse is known as a spin echo. To produce an echo at time TE, the 180° pulse is applied at time TE/2

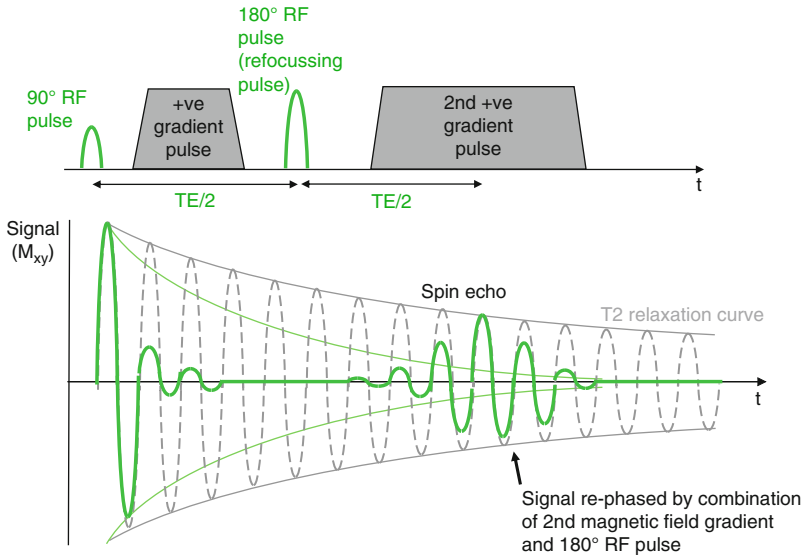

Fig. 5.9 In addition to the effect of the 180° refocusing pulse, gradients are applied to de-phase and re-phase the signal for imaging purposes (see Chap. 6). Note that the second gradient has the same sign as the first as the 180° pulse also changes the sign of the phase shifts caused by the first gradient

is greater than the gradient echo signal. Spin echo images are therefore generally of higher image quality. Imaging based on spin echo is also less affected by the presence of field inhomogeneities caused by metallic artefacts (e.g., sternal wires or metallic heart valves – see also Chap. 17). Gradient echo imaging is however more affected by the presence of magnetic field inhomogeneities caused by iron and so can be useful, for example, in the assessment of patients with increased iron deposition within the heart and liver.

Summary

- Following RF excitation, two relaxation processes occur as the spin system gradually returns back to its equilibrium state.
- As the spin system releases it energy the longitudinal component returns to its equilibrium value through longitudinal relaxation. This is an exponential process with time constant, T1 (also known as T1 relaxation).
- The rotating transverse component decays more rapidly than this due to transverse relaxation. This is also an exponential process and has two causes:
  - Loss of coherence of the rotating proton spins due to the interaction neighbouring magnetic moments (spin-spin interactions). This has a time constant, T2 and is known as T2 relaxation.
  - Loss of coherence between rotating spins due to magnetic field inhomogeneity. This together with T2 relaxation leads to a more rapid decay with a combined time constant, T2*.
- In practice, the FID signal is disrupted by the presence of the magnetic field gradients used to generate images.
- Instead signal echoes are generated and used for the formation of images.
- Gradient echoes are generated by switching the direction of a gradient to first de-phase then re-phase the MR signal
- Spin echoes are generated by applying a 180° RF refocusing pulse to reverse the decay caused by magnetic field inhomogeneities.
- The time from the RF pulse to the maximum amplitude of the echo is known as the echo time, TE.

Further Reading

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