CHAPTER 3

Relaxation

As mentioned in Chapter 2, the MR measurement can be analyzed in terms of energy transfer. The process by which the protons release the energy that they absorbed from the RF pulse is known as relaxation. Relaxation is a fundamental aspect of MR, as essential as energy absorption, and provides the primary mechanism for image contrast as discussed in Chapter 6. In resonance absorption, RF energy is absorbed by the protons only when it is broadcast at the correct frequency. The additional energy disturbs the equilibrium arrangement of spins parallel and antiparallel to $\boldsymbol{B}_0$. Following excitation, relaxation occurs in which the protons release this added energy and return to their original configuration through naturally occurring processes. It is a time-dependent process and is characterized by a rate constant known as the relaxation time. Although it is individual protons that absorb the energy, relaxation times are measured for an entire sample of spins and are statistical or average quantities. Relaxation times are measured for gray matter or cerebrospinal fluid as bulk samples rather than for the individual water or fat molecules within the organs. Two relaxation times can be measured, known as *T1* and *T2*. While both times measure the spontaneous energy transfer by an excited proton, they differ in the final disposition of the energy.

3.1 *T1* relaxation and saturation

The relaxation time *T1* is the time required for the *z* component of $\boldsymbol{M}$ to return to 63% of its original value following an excitation pulse. It is also known as the spin-lattice relaxation time or longitudinal relaxation time. Recall from Chapter 2 that $\boldsymbol{M}_0$ is parallel to $\boldsymbol{B}_0$ at equilibrium and that energy absorption will rotate $\boldsymbol{M}_0$ into the transverse plane. *T1* relaxation provides the mechanism by which the protons give up their energy to return to their original orientation. If a 90° pulse is applied to a sample, $\boldsymbol{M}_0$ will rotate as illustrated in Figure 2.2, and there will be no longitudinal magnetization present following the pulse. As time goes on, a return of the longitudinal magnetization will be observed as

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the protons release their energy (Figure 3.1). This return of magnetization typically follows an exponential growth process, with *T1* being the time constant describing the rate of growth:

$$\boldsymbol{M}(\tau) = \boldsymbol{M}_0(1 - e^{-\tau/T1}) \tag{3.1}$$

where τ is the time following the RF pulse. *T1* relaxation times in tissues are relatively slow, with values ranging from a few milliseconds to several seconds. After 3 *T1* time periods, $\boldsymbol{M}$ will have returned to 95% of its value prior to the excitation pulse, $\boldsymbol{M}_0$. The term spin-lattice refers to the fact that the excited proton ("spin") transfers its energy to its surroundings ("lattice") rather than to another proton. The energy no longer contributes to spin excitation.

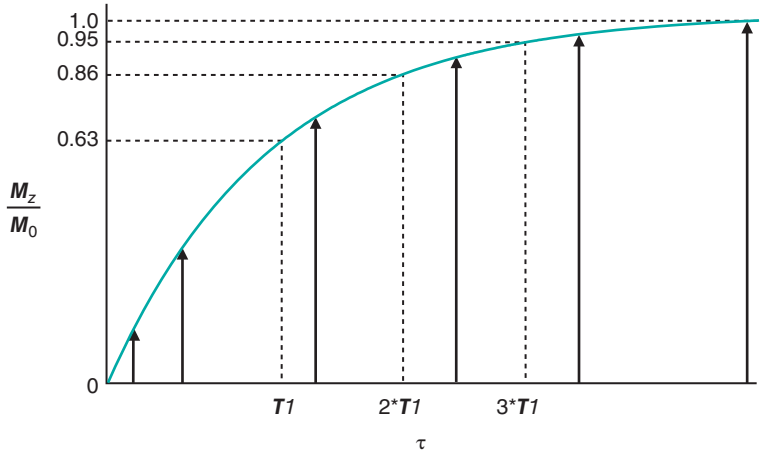

Figure 3.1 *T1* relaxation curve. Immediately following a 90° RF pulse, there is no longitudinal magnetization. A short time later, longitudinal magnetization will be observed as the protons release their energy through *T1* relaxation. Gradually, as more protons release their energy, a larger fraction of $\boldsymbol{M}_Z$ is reestablished. Eventually, $\boldsymbol{M}_0$ will be restored completely. The change of $\boldsymbol{M}_Z/\boldsymbol{M}_0$ with time τ follows an exponential growth process as described by equation (3.1). The time constant for this process is *T1*, the spin-lattice relaxation time, and is the time when $\boldsymbol{M}_Z$ has returned to 63% of its original value.

In a modern MR experiment, pulsed RF energy is applied to the protons repeatedly with a delay time between the pulses. This time between pulses allows the excited protons to give up the absorbed energy (*T1* relaxation). As the protons give up this energy to their surroundings, the population difference (spin up versus spin down) is reestablished so that net absorption can reoccur after the next pulse. In the macroscopic picture, $\boldsymbol{M}$ returns toward its initial value $\boldsymbol{M}_0$ as more energy is dissipated. Since $\boldsymbol{M}$ is the ultimate source of the MR signal, the more energy dissipated, the more signal is generated following the next RF pulse.

For practical reasons, the time between successive RF pulses is usually insufficient for complete *T1* relaxation so that $\boldsymbol{M}$ will not be completely restored to $\boldsymbol{M}_0$. Application of a second RF pulse prior to complete relaxation will rotate $\boldsymbol{M}$ into the transverse plane, but with a smaller magnitude than following the first RF pulse. The following experiment describes the situation (Figure 3.2):

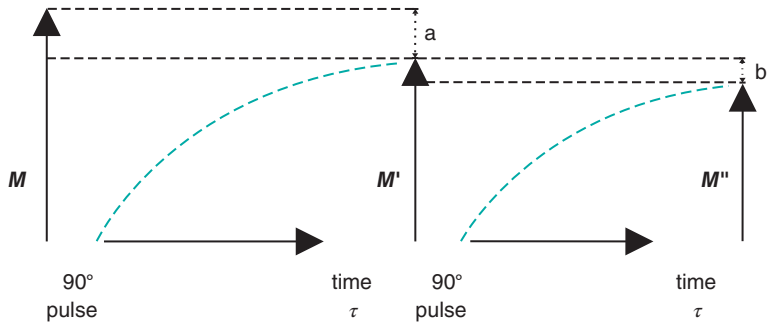

Figure 3.2 Following a 90° RF pulse, longitudinal magnetization is regenerated through *T1* relaxation. If the time between successive RF pulses τ is insufficient for complete recovery of $\boldsymbol{M}$, only $\boldsymbol{M}'$ will be present at the time of the next RF pulse (a). If time τ elapses again, only $\boldsymbol{M}''$ will be present (b). $\boldsymbol{M}''$ will be smaller than $\boldsymbol{M}'$, but the difference will be less than the difference between $\boldsymbol{M}$ and $\boldsymbol{M}'$.

1. A 90° RF pulse is applied. $\boldsymbol{M}$ is rotated into the transverse plane.
2. A time τ elapses, insufficient for complete *T1* relaxation. The longitudinal magnetization at the end of τ, $\boldsymbol{M}'$, is less than in step 1.
3. A second 90° RF pulse is applied. $\boldsymbol{M}'$ is rotated into the transverse plane.
4. After a second time τ elapses, $\boldsymbol{M}''$ is produced. It is smaller in magnitude than $\boldsymbol{M}'$, but the difference is less than the difference between $\boldsymbol{M}$ and $\boldsymbol{M}'$.

Following a few repetitions, $\boldsymbol{M}$ returns to the same magnitude prior to each RF pulse; that is, $\boldsymbol{M}$ achieves a steady-state value. In general, this steady-state value depends on five parameters:

1. The main magnetic field strength B_0. The larger the value for B_0, the larger is M.
2. The number of protons producing $\boldsymbol{M}$ (per unit volume of tissue, known as the proton density).
3. The amount of energy absorbed by the protons (the pulse angle or flip angle).
4. The rate of RF pulse application (time τ).
5. The efficiency of the protons in giving up their energy (*T1* relaxation time).

For many MRI experiments such as standard spin echo and gradient echo imaging, a steady state of $\boldsymbol{M}$ is present because multiple RF pulses are applied and the repetition time *TR* between the pulses is nearly always less than sufficient for complete relaxation. To produce this steady state prior to data collection, additional RF pulses are applied to the tissue immediately prior to the main imaging pulses. These extra RF pulses are known as preparatory pulses or dummy pulses because the generated signals are usually ignored. These preparatory pulses ensure that $\boldsymbol{M}$ has the same magnitude prior to every measurement during the scan.

The rate of RF pulse application and the efficiency of energy transfer must have the proper balance. Suppose the RF energy is applied faster than *T1* relaxation can occur. A comparison of the microscopic and macroscopic pictures is useful at this point. In the microscopic picture, the protons in the lower energy level absorb the RF energy and the protons in the upper energy level are stimulated to emit their energy. As more energy is transmitted, the proton populations of the two levels will gradually equalize. When this equalization occurs, no further net absorption of energy is possible, a condition known as saturation (Figure 3.3). In the macroscopic picture, ***M*** will rotate continuously but gradually get smaller in magnitude until it disappears as the net population difference approaches zero. Since there is no net magnetization, there will be no coherence of proton motion in the transverse plane and thus no signal is produced. This condition is known as saturation. There is a limited amount of energy that a collection of protons can absorb before they become saturated. In a conventional MR measurement, each tissue will experience different amounts of saturation, due to their different *T1* relaxation times. As a result, the contribution of a tissue to ***M*** will differ, as will its maximum potential signal.

As mentioned earlier, spin-lattice relaxation measures the rate of energy transfer from an excited proton to its surroundings. The key to this energy transfer is the presence of some type of molecular motion (e.g., vibration, rotation) of the lattice in the vicinity of the excited proton with an intrinsic frequency, ω_L, that matches the resonant frequency, ω_0. The closer ω_0 is to ω_L, the more readily the motion absorbs the energy and the more frequently this energy transfer occurs, allowing the collection of protons to return to its equilibrium configuration sooner. In tissues, the nature of the protein molecular structure and any metal ions that may be present have a pronounced effect on the particular ω_L. Metals ions such as iron or manganese can have significant magnetic moments that may influence the local environment. While the particular protein structures are different for many tissues, the molecular rotation or tumbling of most proteins typically have ω_L of approximately 1 MHz. Therefore, at lower resonant frequencies (lower B_0), there is a better match

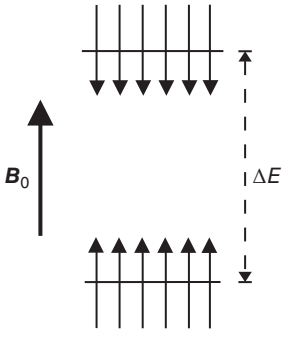

Figure 3.3 Saturation. If RF pulses are applied faster than the *T1* relaxation processes can dissipate the energy, the spin populations equalize between the two energy levels. As a result, there is no difference in the number of spins and no net magnetization.

between ω_L and ω_0. This enables a more efficient energy transfer to occur and thus *T1* is shorter. This is the basis for the frequency dependence of *T1*; namely, *T1* decreases with decreasing strength of the magnetic field. This is also the reason that a larger B_0 does not necessarily translate to a greater signal, as saturation is more prevalent due to the longer *T1* times.

3.2 *T2* relaxation, *T2** relaxation, and spin echoes

The relaxation time *T2* is the time required for the transverse component of $\boldsymbol{M}$ to decay to 37% of its initial value through irreversible processes. It is also known as the spin–spin relaxation time or transverse relaxation time. Recall from Chapter 1 that $\boldsymbol{M}_0$ is oriented only along the *z* ($\boldsymbol{B}_0$) axis at equilibrium and that no portion of $\boldsymbol{M}_0$ is in the *xy* plane. The coherence is entirely longitudinal or parallel to $\boldsymbol{B}_0$. Absorption of energy from a 90° RF pulse as in Figure 2.2 causes $\boldsymbol{M}_0$ to rotate entirely into the *xy* plane, so that the coherence is in the transverse plane at the end of the pulse. As time elapses, this coherence disappears while at the same time the protons release their energy and reorient themselves along $\boldsymbol{B}_0$. This disappearing coherence produces the FID described in Chapter 2. As this coherence disappears, the value of $\boldsymbol{M}$ in the *xy* plane decreases toward 0. *T2* or *T2** relaxation is the process by which this transverse magnetization is lost.

A comparison of the microscopic and macroscopic pictures provides additional insight. At the end of the 90° RF pulse, when the protons have absorbed energy and $\boldsymbol{M}$ is oriented in the transverse plane, each proton precesses at the same frequency ω_0 and is synchronized at the same point or phase of its precessional cycle. Since a nearby proton of the same type will have the same molecular environment and the same ω_0, it can readily absorb the energy that is being released by its neighbor. Spin–spin relaxation refers to this energy transfer from an excited proton to another nearby proton (Figure 3.4). The absorbed energy remains as spin excitation

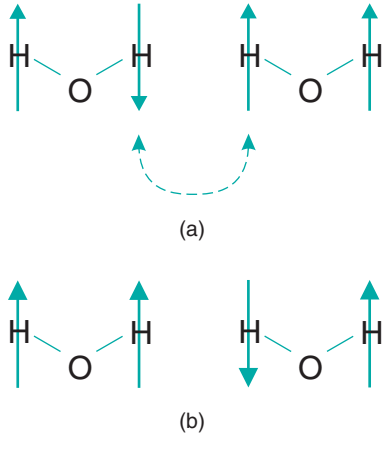

Figure 3.4 Spin– spin relaxation. (a) Two water molecules, with one hydrogen atom on one molecule having absorbed RF energy and being excited (spin down). (b) If the molecules are in close proximity, the energy can be transferred from the first water molecule to a hydrogen atom on the second water molecule.

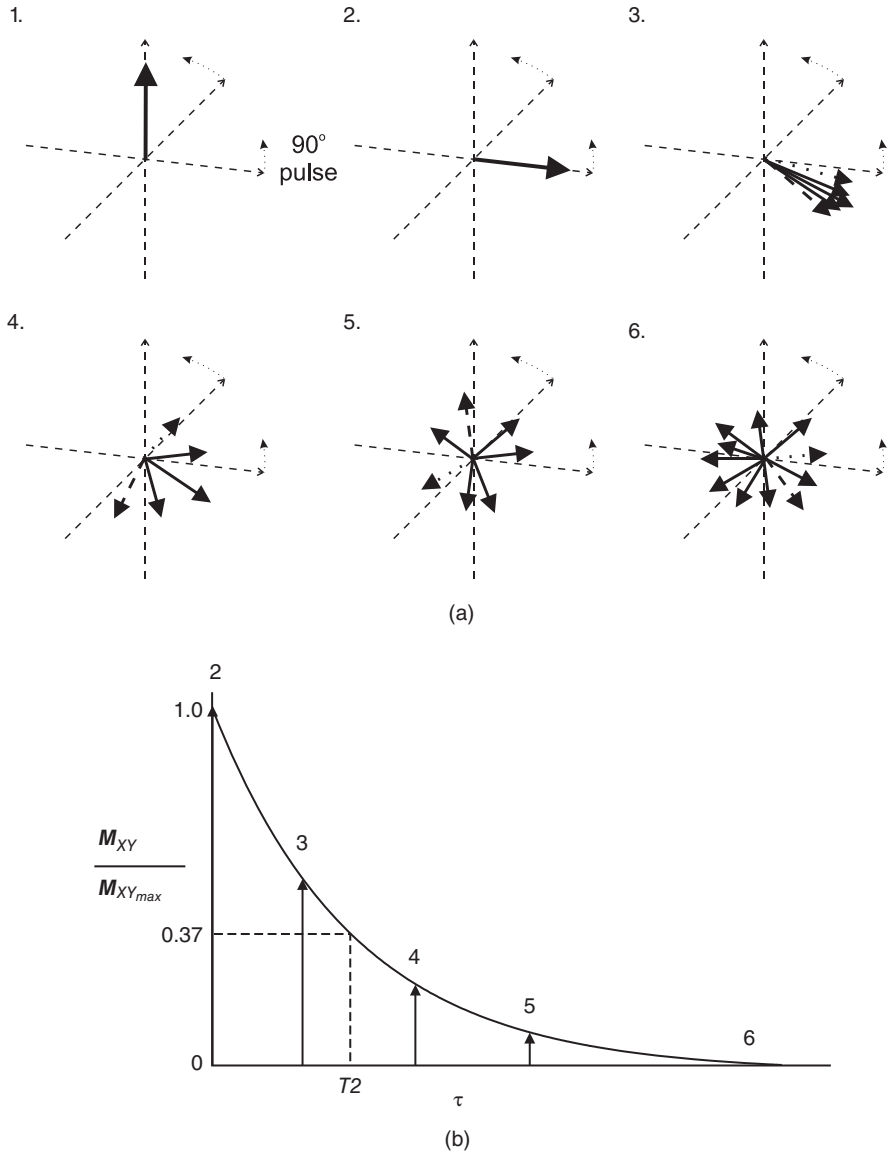

Figure 3.5 (a) A rotating frame slower than ω_0 is assumed for this figure. Net magnetization ***M*** (arrow) is oriented parallel to $\boldsymbol{B}_0$ (not shown) prior to an RF pulse (1). Following a 90° RF pulse, the spins precess initially in phase in the transverse plane (2). Due to inter-and intramolecular interactions, the spins begin to precess at different frequencies (dashed arrow, faster; dotted arrow, slower) and become asynchronous with each other (3). As more time elapses (4,5), the transverse coherence becomes smaller until there is complete randomness of the transverse components and no coherence (6). (b) Plot of relative $\boldsymbol{M}_{XY}$ component as a function of time. The numbers correspond to the expected $\boldsymbol{M}_{XY}$ component from (a). The change in $\boldsymbol{M}_{XY}/\boldsymbol{M}_{XY\,max}$ with time follows an exponential decay process as described by equation (3.3). The time constant for this process is the spin-spin relaxation time *T2** and is the time when $\boldsymbol{M}_{XY}$ has decayed to 37% of its original value.

rather than being transferred to the surroundings as in *T1* relaxation. This proton–proton energy transfer can occur many times as long as the protons are in close proximity and remain at the same ω_0. Intermolecular and intramolecular interactions such as vibrations or rotations cause transient fluctuations in the magnetic field and thus cause ω_0 to fluctuate. This fluctuation produces a gradual, irreversible loss of phase coherence to the spins as they exchange the energy and reduce the magnitude of the transverse magnetization and the generated signal (Figure 3.5). *T2* is the time when the transverse magnetization is 37% of its value immediately after the 90° pulse when this irreversible process is the only cause for the loss of coherence. As more time elapses, this transverse coherence completely disappears, only to reform in the longitudinal direction as *T1* relaxation occurs. This dephasing time *T2* is always less than or equal to *T1*.

There are several potential causes for a loss of transverse coherence to $\boldsymbol{M}$. One is the molecular motions of the adjacent spins due to vibrations or rotations. This irreversible movement is responsible for spin–spin relaxation or the true *T2*. Another cause arises from the fact that a proton never experiences a magnetic field that is 100% uniform or homogeneous in value. As the proton precesses, it experiences a fluctuating local magnetic field, causing a change in ω_0 and a loss in transverse phase coherence. This nonuniformity in $\boldsymbol{B}_0$ comes from three sources:

1. Main field inhomogeneity. There is always some degree of nonuniformity to $\boldsymbol{B}_0$ due to imperfections in magnet manufacturing, interactions with nearby building walls, or other sources of metal. This field distortion is constant during the measurement time.
2. Sample-induced inhomogeneity. Differences in the magnetic susceptibility or degree of magnetic polarization of adjacent tissues (e.g., bone, air) will distort the local magnetic field near the interface between the different tissues. Provided there is no motion of the sample, this inhomogeneity is also of constant magnitude and is present as long as the patient is present within the magnet.
3. Imaging gradients. As discussed in Chapter 4, the technique used for spatial localization generates a magnetic field inhomogeneity that induces proton dephasing. This inhomogeneity is transient during the measurement.

The contributions of the imaging gradients may be eliminated as a source of dephasing through proper design of the measurement process, as described in Chapter 4. The other sources contribute to the total transverse relaxation time, *T2**:

$$1/T2^* = 1/T2 + 1/T2_{\mathrm{M}} + 1/T2_{\mathrm{MS}} \tag{3.2}$$

where $T2_{\mathrm{M}}$ is the dephasing time due to the main field inhomogeneity and $T2_{\mathrm{MS}}$ is the dephasing time due to the magnetic susceptibility differences. The decay of the transverse

magnetization following a 90° RF pulse, the FID, follows an approximately exponential process with the time constant of $T2^*$ rather than just $T2$:

$$\boldsymbol{M}_{XY}(t) = \boldsymbol{M}_{XY\,\max} e^{-t/T2^*} \tag{3.3}$$

where $\boldsymbol{M}_{XY\,\max}$ is the transverse magnetization $\boldsymbol{M}_{XY}$ immediately following the excitation pulse. For most tissues or liquids, $T2_M$ is the major factor in determining $T2^*$, while for tissue with significant iron deposits or air-filled cavities, $T2_{MS}$ dominates $T2^*$.

Some sources of proton dephasing can be reversed by the application of a 180° RF pulse, which is described by the following sequence of events (Figure 3.6):

1. A 90° RF pulse.
2. A short delay of time t.
3. A 180° RF pulse.
4. A second time delay t.

The initial 90° RF pulse rotates $\boldsymbol{M}_0$ into the transverse plane. During the time t, proton dephasing will occur through $T2^*$ relaxation processes and the transverse coherence will diminish. Application of the 180° RF pulse causes the protons to reverse their phases relative to the resonant frequency. The rates and directions of precession for the protons do not change, only their relative phase. If time t elapses again, then the protons will regain some of their transverse coherence. This reformation of phase coherence induces another signal in the receiver coil, known as a spin echo. Sources of dephasing that do not change during the two time periods, the main field inhomogeneity and magnetic susceptibility differences, are eliminated because the protons experience exactly the same interactions prior to and following the 180° pulse. This means that the contributions to $T2^*$ relaxation from these static sources will disappear. Only the irreversible spin–spin relaxation is not recovered by the 180° RF pulse so that the loss of phase coherence and signal amplitude for a spin echo is due only to true $T2$ relaxation.

Following the echo formation, the protons continue to precess and dephase a second time as the sources of dephasing continue to affect them. Application of a second 180° RF pulse again reverses the proton phases and generates another coherence to the protons, producing another spin echo. This second echo differs from the first echo by the increased amount of $T2$ relaxation contributing to the signal loss. This process of spin echo formation by 180° RF pulses can be repeated as many times as desired, until $T2$ relaxation completely dephases the protons. The use of multiple 180° pulses maintains phase coherence to the protons longer than the use of a single 180° RF pulse because of the significant dephasing that the field inhomogeneity induces over very short time periods.

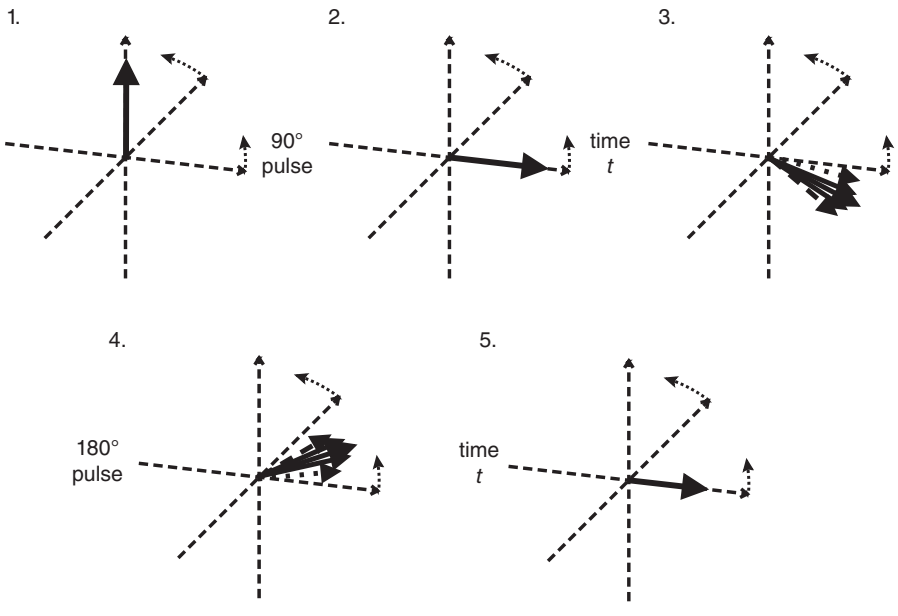

Figure 3.6 A rotation frame slower than ω_0 is assumed for this figure. Net magnetization $\boldsymbol{M}$ (arrow) is oriented parallel to $\boldsymbol{B}_0$ (not shown) prior to an RF pulse (1). Application of a 90° RF pulse rotates $\boldsymbol{M}$ into the transverse plane (2). Due to the $T2^*$ relaxation processes, the protons become asynchronous with each other during time t (3). Application of a 180° RF pulse causes the protons to reverse their phase relative to the transmitter phase. The protons that precessed most rapidly are farthest behind (dashed arrow), while the slowest protons are in front (dotted arrow) (4). Allowing time t to elapse again allows the protons to regain their phase coherence in the transverse plane (5), generating a signal in the receiver coil known as a spin echo. The loss in magnitude of the reformed coherence relative to the original coherence (2) is due to irreversible processes (i.e., true spin–spin or $T2$ relaxation). Equation (3.3) describes the decay of $\boldsymbol{M}_{XY}$ if $T2$ is used instead of $T2^*$.

One important difference between $T1$ and $T2$ relaxation is in the influence of B_0. As mentioned earlier, $T1$ is very sensitive to B_0, with longer $T1$ times measured for a tissue at higher B_0. $T2$ is relatively insensitive to B_0 at the relatively large field strengths currently used in MRI. Only at very low B_0 (less than 0.05 T) will there be significant changes in $T2$. The other components of $T2^*$, $T2_M$ and $T2_{MS}$ become more prominent at higher B_0. Good magnetic field uniformity is more difficult to generate at high magnetic fields, so that $T2_M$ will be shorter. Greater B_0 will also cause greater differences in M_0 between two tissues with different magnetic susceptibilities, producing shorter $T2_{MS}$. The result is that $T2$-weighted techniques will see little sensitivity to B_0 while $T2^*$-weighted techniques will show greater signal differences at higher B_0.