Lecture Three

MRI DESIGN: Relaxation Mechanisms

Relaxation is a general term that refers to a loss of energy. In MRI this is energy delivered to the spins via excitation. After the RF excitation pulse has been applied and resonance and the desired flip angle achieved, the RF pulse is removed. The signal induced in the receiver coil begins to decrease. This is because the coherent component of NMV in the transverse plane, which is passing across the receiver coil, begins to gradually decrease as an increasingly higher proportion of spins become out of phase with each other. The amplitude of the voltage induced in the receiver coil therefore gradually decreases. This is called **free induction decay** or **FID**. The NMV in the transverse plane decreases due to:

- relaxation processes;
- field inhomogeneities and susceptibility effects.

The cumulative dephasing of spin-spin interactions and inhomogeneities is called T2* decay (Table 7.1).

Table 7.1 Common equations of relaxation mechanisms.

Equations		
$1/T2^* = 1/T2 + 1/2\ \gamma\Delta B_0$	T2 and T2* are the tissues' T2 and T2* relaxation times (ms) γ is the gyromagnetic ratio (MHz/T) ΔB_0 is the variation in magnetic field (inhomogeneities) (parts per million, ppm)	This equation shows how T2 and T2* are related to each other. Poor field inhomogeneities result in T2* being much shorter than T2, and fast decaying signal.

Relaxation processes

The magnetization in each tissue relaxes at different rates. This is one of the factors that create image contrast. The withdrawal of the RF produces several effects:

• Spins emit energy absorbed from the RF pulse through a process known as **spin lattice energy transfer** and shift their magnetic moments from the *high-energy state* to the *low-energy state*. The NMV recovers and realigns to B_0. This relaxation process is called ***T1 recovery***.

• The magnetic moments of the spins lose precessional coherence or dephase and the *NMV decays in the transverse plane*. The dephasing relaxation process is called ***T2 decay***.

The magnetic moments of the spins lose their coherence by:

• interactions of the intrinsic magnetic fields of adjacent nuclei (**spin-spin**) causing **T2 decay**;

• **inhomogeneities** of the external magnetic field.

Field inhomogeneities

Despite attempts to make the main magnetic field as uniform as possible via shimming, inhomogeneities of the external magnetic field are inevitable and slightly alter the magnitude of B_0; that is, some small areas of the field have a magnetic field strength of slightly more or less than the main field strength.

Due to the Larmor equation, the precessional frequency of the magnetic moment of a spin is proportional to B_0, so spins that pass through inhomogeneities experience a precessional frequency and phase change, and the resulting signal decays exponentially. This results in a change in dephasing of the transverse magnetization due to a loss in phase coherence (Figure 7.1). The resulting signal decays exponentially and is called an FID.

T2 decay is irreversible because spin-spin interactions occur at the atomic or molecular level. However, T2* decay, particularly caused by field inhomogeneity, can be compensated for and is desirable (Table 7.1). In order to produce images where T2 contrast is visualized, ideally there must be a mechanism to rephase spins and compensate for magnetic field inhomogeneities. *Pulse sequences* are mechanisms that perform this function.

A **pulse sequence** is defined as a series of RF pulses, gradient applications, and intervening time periods. They enable control of the way in which the system applies RF pulses and gradients. By selecting the intervening time periods, image weighting is controlled. Pulse sequences are required because, without a mechanism of refocusing spins, there is an insufficient signal to produce an image. This is because dephasing occurs almost immediately after the RF excitation pulse has been removed.

The main purposes of pulse sequences are to:

- rephase spins and remove inhomogeneity effects and therefore produce a signal or echo that contains information about the T2 decay characteristics of tissue alone;
- enable manipulation of the TE and TR to produce different types of contrast.

Spins are rephased by using (Table 7.2):

- a 180° RF pulse (used in all spin echo sequences);
- a gradient (used in all gradient echo sequences).

Table 7.2 Pulse sequences and their rephasing mechanisms.

Use RF pulses to rephase spins	Use gradients to rephase spins
Conventional spin echo	Coherent gradient echo
Fast or turbo spin echo	Incoherent gradient echo
Inversion recovery	Steady-state free precession
STIR	Balanced gradient echo
FLAIR	EPI

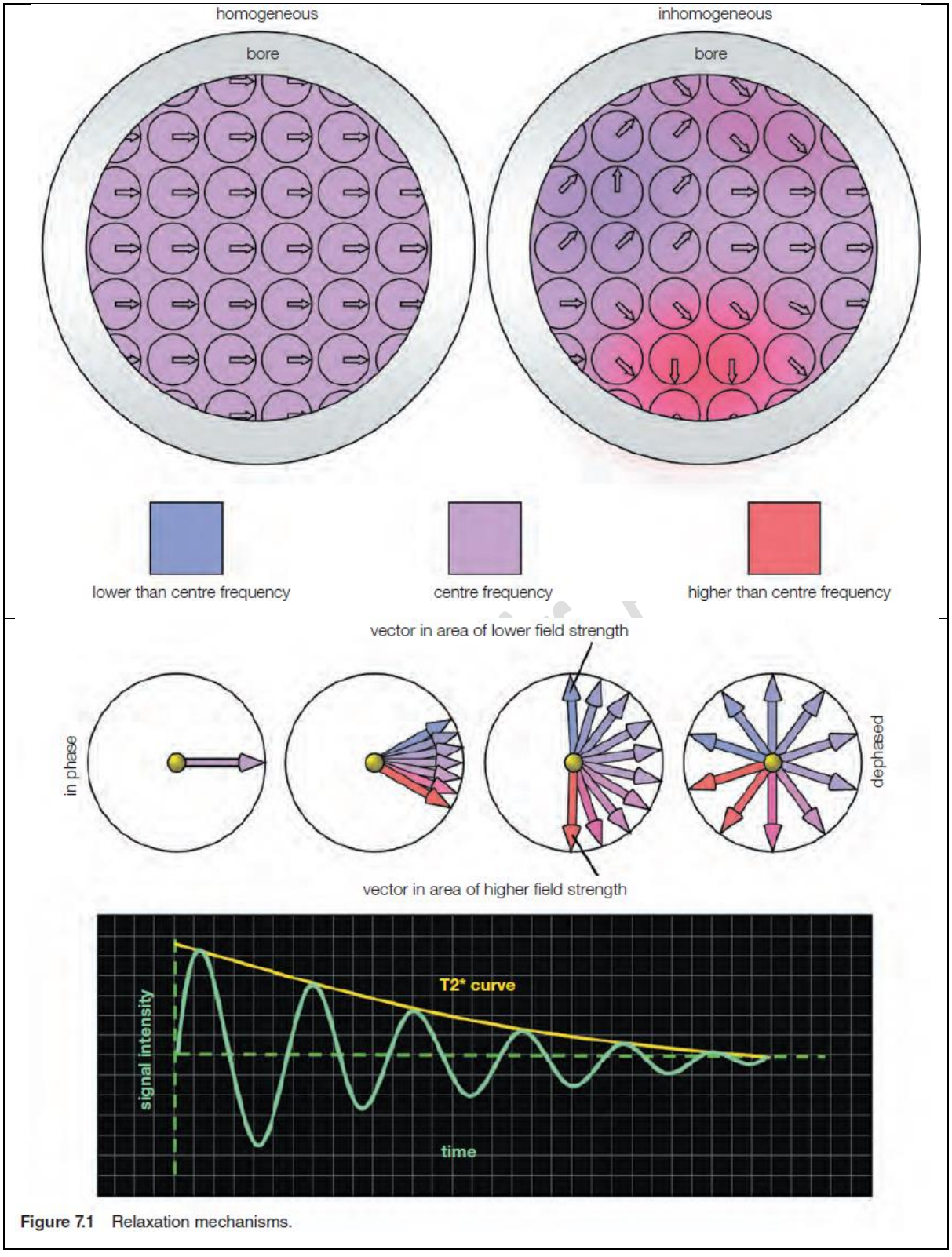


Figure 7.1 Relaxation mechanisms.

[Q: What is RF pulse in MRI? And what is gradient in MRI?]

KEY POINTS

✓ Relaxation is a general term that refers to a loss of energy. In MRI, this is energy delivered to the spins via excitation.

✓ Relaxation and inhomogeneities result in a FID signal.

✓ Spin lattice energy transfer is a relaxation process where spins give up the energy absorbed through excitation to the surrounding molecular lattice of the tissue. It is called T1 recovery.

✓ T2 decay is an irreversible loss of phase coherence due to spin-spin interactions on an atomic and molecular level.

✓ Pulse sequences are mechanisms that permit refocusing of spins so that images can be acquired with different types of contrast.

MRI DESIGN: T1 Recovery

T1 recovery is caused by the exchange of energy from spins to their surrounding environment or lattice. It is called **spin lattice energy transfer**. As the spins dissipate their energy their magnetic moments relax or return to B_0; that is, they regain their longitudinal magnetization. Th e rate at which this occurs is an exponential process and it takes place at different rates in different tissues.

The T1 recovery time of a particular tissue is an intrinsic contrast parameter that is inherent to the tissue being imaged. It is a constant for a particular tissue and is defined as the time it takes for 63% of the longitudinal magnetization to recover in that tissue (Figure 8.1 and Table 8.1). The period of time during which this occurs is the time between one excitation pulse and the next or the **TR**. The TR therefore determines how much T1 recovery occurs in a particular tissue.

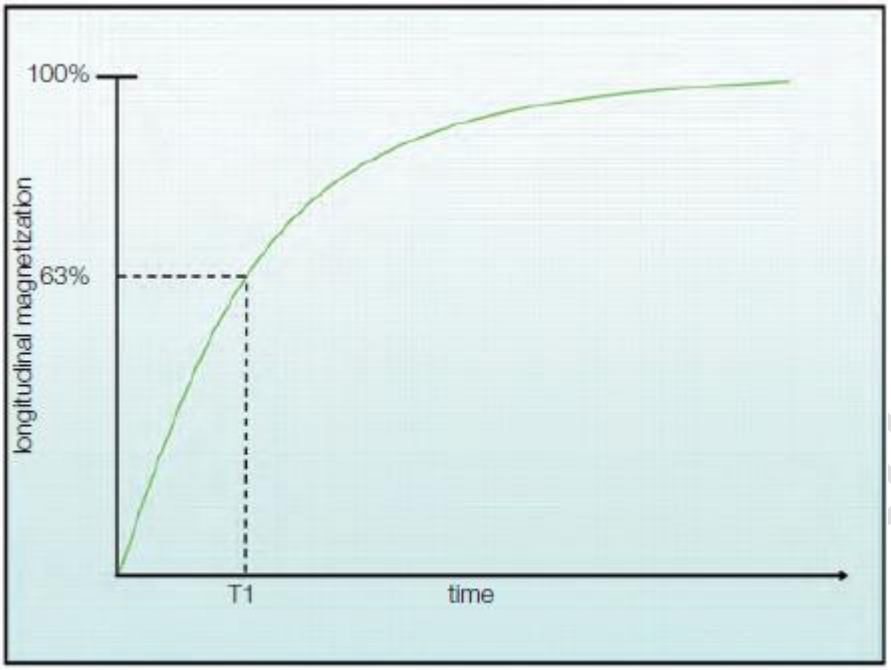


Figure 8.1 The T1 recovery curve.

Table 8.1 Equations of T1 recovery.

Equations		
$Mz_t = Mz\,(1-e^{-t/T1})$	Mz_t is the amount of longitudinal magnetization at time t after the removal of the excitation pulse. Mz is full longitudinal magnetization. T1 is the T1 recovery time (ms) and is the time taken to increase the longitudinal magnetization by a factor of e.	This equation plots the size of the recovering NMV as a function of time after the removal of the excitation pulse and the T1 recovery time. When t=T1, 63% of the longitudinal magnetization recovers. When t=2xT1, 86% recovers and when t=3xT1, 95% recovers. It usually takes between 3 and 5 T1 recovery times for full recovery to occur.

T1 recovery in fat

T1 relaxation occurs as a result of spins exchanging the energy given to them by the RF pulse to their surrounding environment. The efficiency of this process determines the T1 recovery time of the tissue in which they are situated. Due to the fact that fat is able to absorb energy quickly, the *T1 recovery time of fat is very short*; that is, spins dispose of their energy to the surrounding fat tissue and return to B_0 in a short time (Figure 8.2; Table 8.2).

Table 8.2 T1 relaxation times of brain tissue at 1 T.

Tissue	T1 relaxation time (ms)
Water	2500
Fat	200
CSF	2000
White matter	500

T1 recovery in water

Water is very inefficient at receiving energy from spins. *The T1 recovery time of water is therefore quite long*; that is, spins take a lot longer to dispose of their energy to the surrounding water tissue and return to B_0 (Figure 8.3; Table 8.2). In addition, the efficiency of spin lattice energy transfer depends on how closely molecular motion of the molecules matches the Larmor frequency. If there is a good match between the rate of molecular tumbling and the precessional frequency of spins, energy is efficiently exchanged between hydrogen and the surrounding molecular lattice. The Larmor frequency is relatively slow and therefore fat is much better at this type of energy exchange than water, whose molecular motion is much faster than the Larmor frequency. *This is another reason why fat has a shorter T1 recovery time than water.*

T1 recovery is affected by the strength of the external magnetic field. The precessional frequency of spins within a tissue varies slightly, but efficient energy exchange due to molecular motion only occurs at the Larmor frequency. The Larmor frequency is proportional to B_0 and therefore T1 recovery takes longer as B_0 increases, because there are fewer molecules moving at relaxation-causing frequencies.

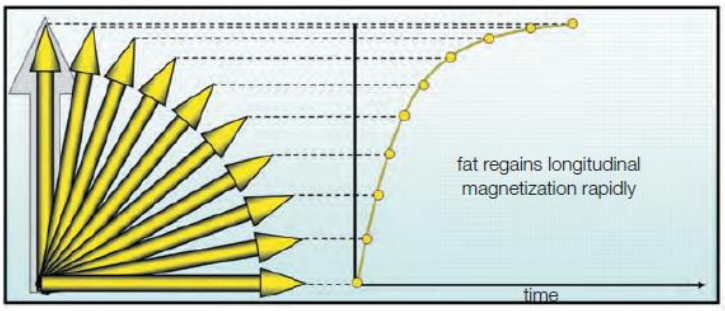


Figure 8.2 T1 recovery in fat.

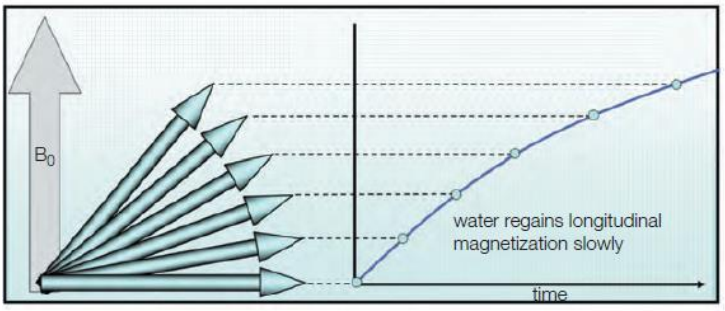


Figure 8.3 T1 recovery in water.

Control of T1 recovery

The TR controls how much of the NMV in fat or water recovers before the application of the next RF pulse. *A short TR* does not permit full longitudinal recovery in most tissues, so that there are different longitudinal components in fat and water. These different longitudinal components are converted to different transverse components after the next excitation pulse has been applied.

As the NMV does not recover completely to the positive longitudinal axis, they are pushed beyond the transverse plane by the succeeding 90° RF pulse. This is called **saturation**. When saturation occurs, there is a contrast difference between fat and water due to differences in their T1 recovery times (Figures 8.4 and 8.5).

A long TR allows full recovery of the longitudinal components in most tissues. There is no difference in the magnitude of their longitudinal components.

There is no contrast difference between fat and water due to differences in T1 recovery times when using a long TR. Any differences seen in contrast are due to differences in the number of protons or **proton density** of each tissue. The proton density of a particular tissue is an intrinsic contrast parameter and is therefore inherent to the tissue being imaged.

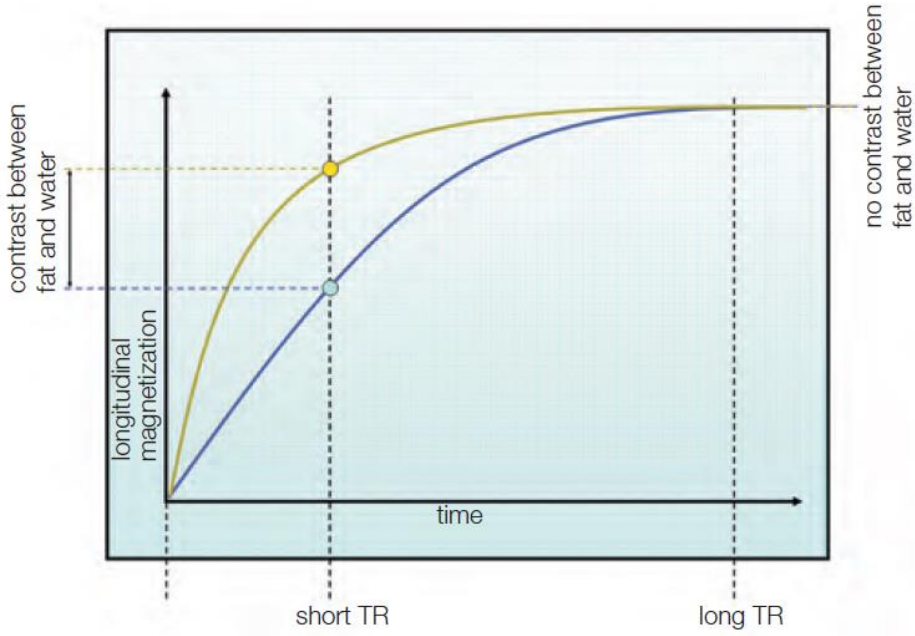


Figure 8.4 T1 recovery of fat and water.

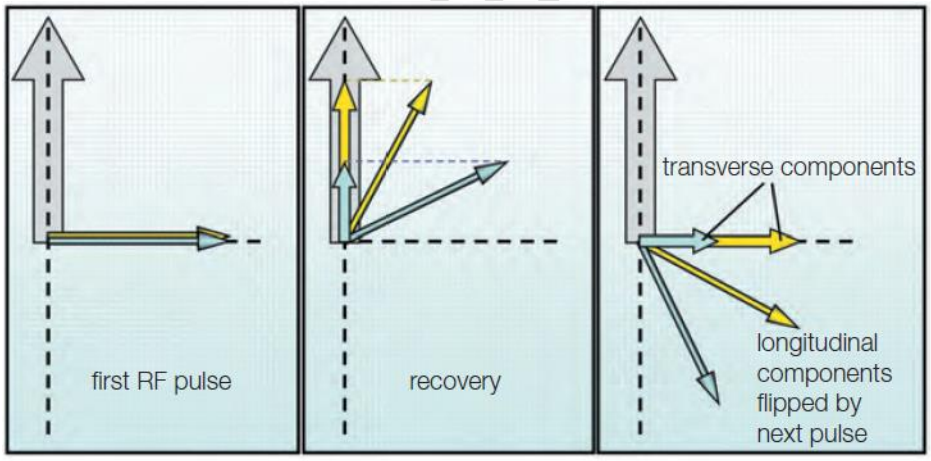


Figure 8.5 T1 contrast generation.

KEY POINTS

- ✓ Fat has a short T1 recovery time.
- ✓ Water has a long T1 recovery time.
- ✓ T1 recovery is caused by spin lattice energy transfer. The efficiency of this process depends on the inherent energy of the tissue and how well the rate of molecular tumbling matches Larmor.
- ✓ T1 recovery times are dependent on magnetic field strength. As field strength increases, tissues take longer to relax.
- ✓ T1 contrast is controlled by the TR. For good T contrast, the TR must be short.

MRI DESIGN: T2 Decay

T2 decay is caused by the interaction between the magnetic fields of neighbouring spins. It is called **spin-spin**. It occurs as a result of the intrinsic magnetic fields of the nuclei interacting with each other. This produces a loss of phase coherence or dephasing, and results in decay of the NMV in the transverse plane. It is an exponential process and occurs at different rates in different tissues.

The T2 decay time of a particular tissue is an intrinsic contrast parameter and is inherent to the tissue being imaged. It is the time it takes for 63% of the transverse magnetization to be lost due to dephasing; that is, transverse magnetization is reduced by 63% of its original value (37% remains; Figure 9.1 and Table 9.1). The period of time during which this occurs is the time between the excitation pulse and the MR signal or the **TE**. The TE therefore determines how much T2 decay occurs in a particular tissue.

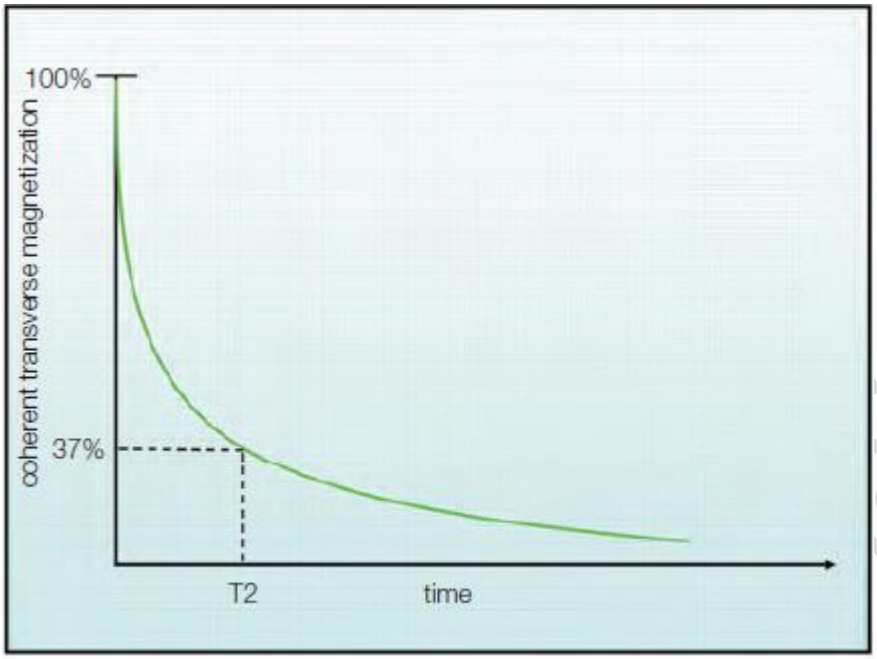


Figure 9.1 The T2 decay curve.

Table 9.1 T2 decay times of brain tissue at 1 T.

Equations		
$Mxy_t = Mxy\ e^{-t/T2}$	Mxy_t is the amount of transverse magnetization at time t (ms) after the removal of the excitation pulse. Mxy is full transverse magnetization. T2 is the T2 decay time (in ms) and is the time taken to reduce the transverse magnetization by a factor of e.	This equation plots the size of the decaying transverse magnetization as a function of time after the removal of the excitation pulse and the T2 decay time. When t=T2, 63% of the coherent transverse magnetization has decayed and 37% remains.

T2 decay in fat

T2 relaxation occurs as a result of the spins of adjacent nuclei interacting with each other. The efficiency of this process depends on how closely packed the molecules are to each other. In fat, the molecules are closely packed together so that spin-spin is efficient. *The T2 time of fat is therefore very short* (Figure 9.2).

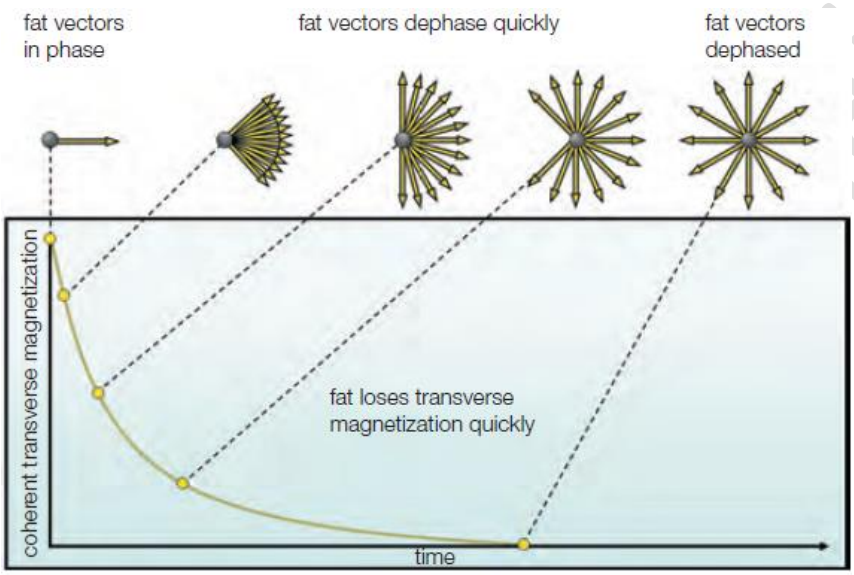


Figure 9.2 T2 decay in fat.

T2 decay in water

In water the molecules are spaced apart so that spin-spin is not efficient. *The T2 time of water is therefore very long* (Figure 9.3).

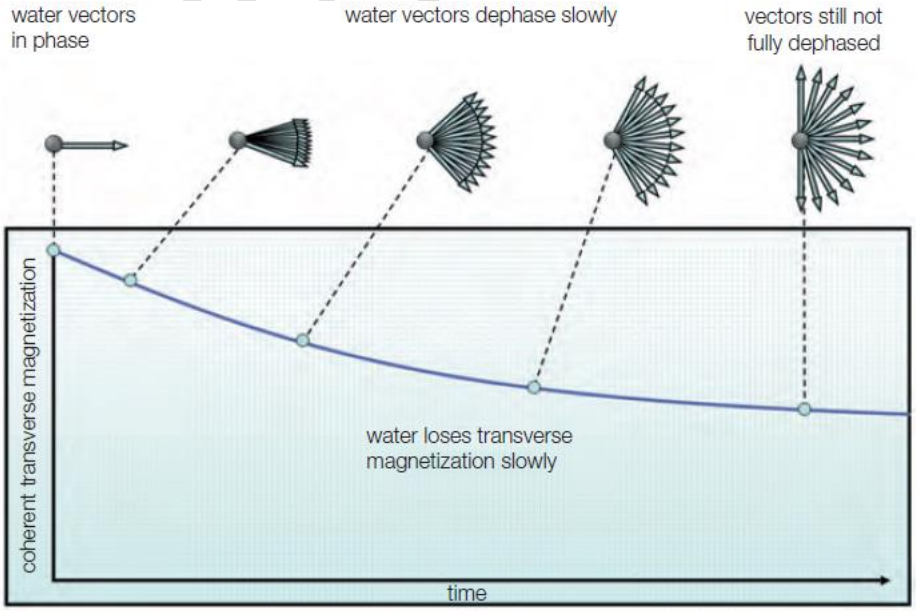


Figure 9.3 T2 decay in water.

Control of T2 decay

The **TE** controls how much transverse magnetization has been allowed to decay in fat and water when the signal is read.

A *short TE* does not permit full dephasing in either fat or water, so their coherent transverse components are similar. There is little contrast difference between fat and water due to differences in T2 decay times using a short TE.

A *long TE* allows dephasing of the transverse components in fat and water. There is a contrast difference between fat and water due to differences in T2 decay times when using a long TE (Figure 9.4).

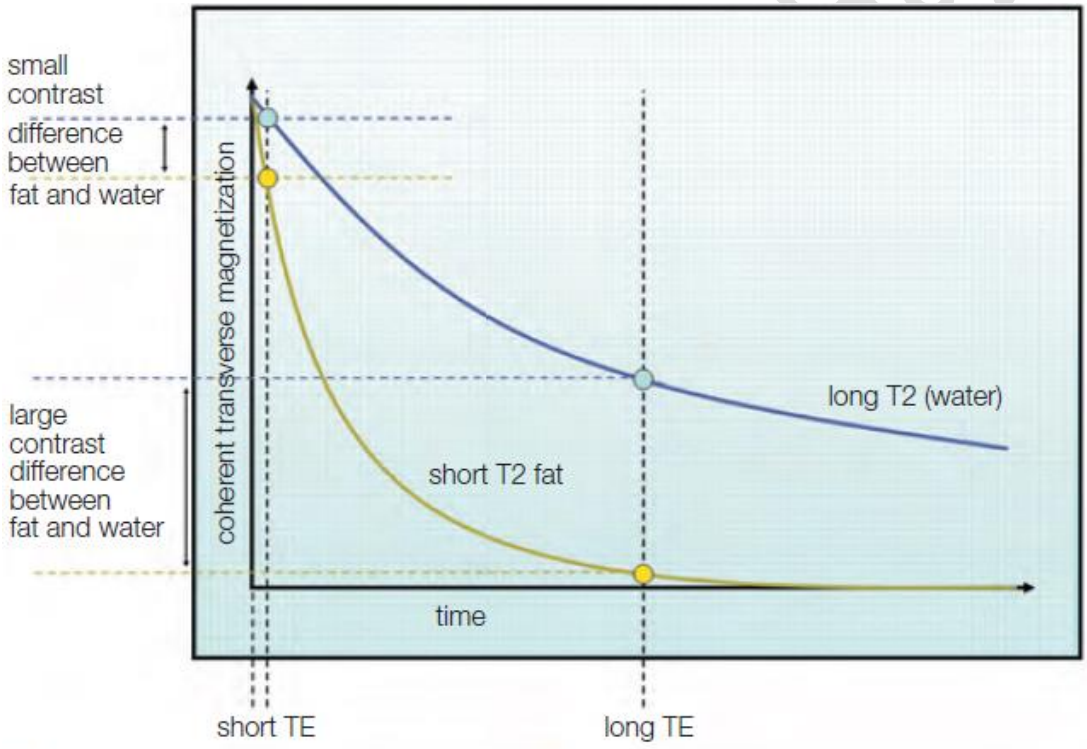


Figure 9.4 T2 decay curves in fat and water.

T2 decay is affected by the strength of the external magnetic field. Spin-spin processes are more efficient when molecular motion occurs at the Larmor frequency. The Larmor frequency is proportional to B_0 and therefore T2 decay takes longer as B_0 increases, because there are fewer molecules moving at relaxation-causing frequencies. It should be noted that fat and water represent the extremes in image contrast. Other tissues, such as muscle, grey matter and white matter, have contrast characteristics that fall between fat and water (Table 9.2).

Table 9.2 Equations of T2 decay.

Tissue	T2 decay time (ms)
Water	2500
Fat	100
CSF	300
White matter	100

Notes:

The **stationary frame** of reference refers to the observer (i.e. you) viewing something moving. You and the room you are situated in are stationary and what you are observing moves.

The **rotating frame of reference** refers to the observer viewing this from a different perspective. Imagine you are the thing that moves, what would the room look like? You would appear stationary and the room would appear to move.

A good example of this is to imagine the NMV relaxing back to B_0. If you were to observe this from the stationary frame of reference, as the NMV relaxes it also precesses around B_0. If you are looking at this from the rotating frame, however, *you* become the NMV as if you 'ride along' with it. From this perspective the room moves around you (the NMV) and you just smoothly relax back to B_0. In other words, from the rotating frame of reference it is the room that precesses relative to the NMV, rather than the NMV precessing relative to the room (as with the stationary frame of reference).

KEY POINTS

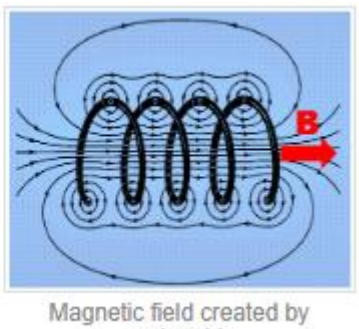
- ✓ Fat has a short T2 decay time.
- ✓ Water has a long T2 decay time.
- ✓ T2 decay is caused by spin-spin energy transfer. The efficiency of this process depends on how closely the molecules are packed together.
- ✓ T2 decay times are dependent on magnetic field strength. As field strength increases, tissues take longer to dephase.
- ✓ T2 contrast is controlled by the TE. For good T2 contrast, the TE must be long.

MRI DESIGN: Magnets

MR scanners are broadly classified into closed- and open-bore systems. Over 90% of scanners worldwide are of the closed-bore cylindrical design and generate their fields by passing current through a solenoid kept at superconducting temperatures. Open bore magnets contain an air gap between two magnetic poles. These may utilize permanent magnets or electromagnets.

Closed bore (cylindrical) configuration with superconducting solenoidal design. The coils are bathed in liquid helium allowing a stable, homogeneous field to be created, typically 1T and higher.

GE Signa 1.5T superconducting scanner

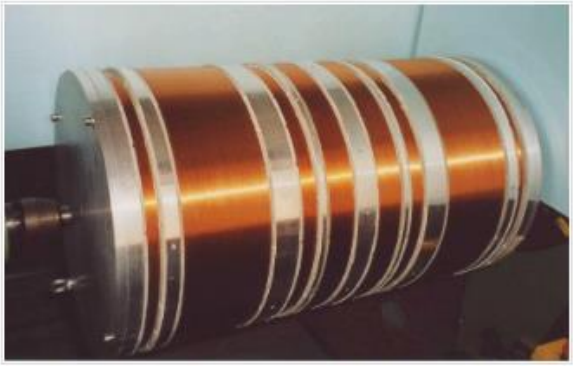


Magnetic field created by solenoid

The design of cylindrical magnets is not quite as simple as pictured above. A single continuous solenoidal winding will only produce a completely homogenous central field if infinite in length. For coils truncated to the 2-3 meters range, a single solenoid would not provide a sufficiently uniform field for

MR imaging. The homogeneity problem is solved by breaking up the main coil into 6-10 separate windings with gaps in between as shown below.

This configuration maintains symmetry of the field along the z-axis (B_o direction), minimizes fringe fields at the ends of the scanner, and improves homogeneity centrally. There are, as expected, some minor fluctuations in field strength along the z-axis, with minimally higher fields directly under the bands and minimally lower fields within the gaps.

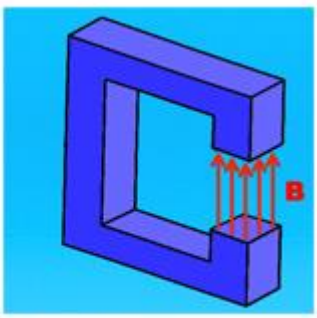


Solenoidal superconducting magnet under construction before being placed in cryostat.

Rather than a single large/continuous winding, 6-10 large and small solenoidal bands are typically placed to create a more uniform central field with minimized fringe.

Most ***open bore*** scanners utilize permanent magnets in a C-shaped or horseshoe configuration. These operate at field strengths typically ranging from 0.064T to 1.0T.

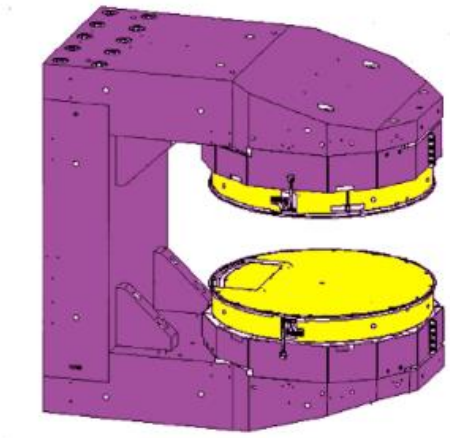
Hitachi Aperto 0.4T permanent magnet scanner



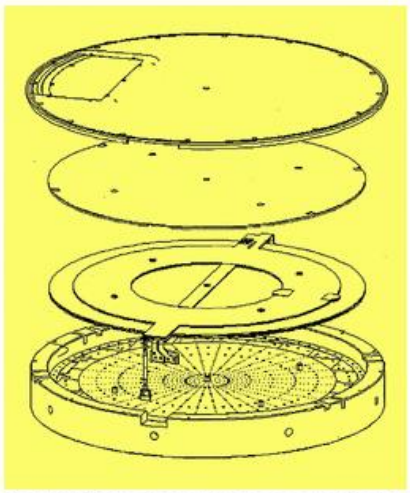
C-shaped permanent magnet

Although it is tempting to think of the entire C-shaped structure being a solid permanent magnet, that is not actually the case. C-shaped open magnets are composed of a ferromagnetic yoke and pole pieces (shoes). The yoke itself is not magnetically saturated. Field generation is created by a pair of magnets located at the ends of the yoke on both sides of the air gap.

For permanent magnet scanners, these pole pieces are often a set of *neodymium-boron-iron magnetic* blocks or disks. Resistive or superconducting electromagnets can also be used. In the latter case, the yoke is not a flux link but only serves to hold the superconducting magnets in exact alignment.



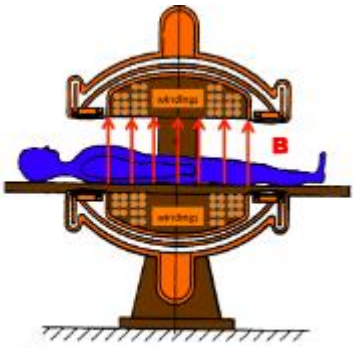
C-shaped open magnet scanner showing iron yoke (magenta) with pole pieces (yellow).



Exploded view of pole piece showing an array of neodymium-boron-iron magnetic blocks.

The third design is a ***dipolar electromagnet configuration*** with coils on either side of the patient. These coils can be superconductive or resistive and range from 0.5T to 1.2T.

Hitachi Oasis 1.2T HFO superconducting scanner

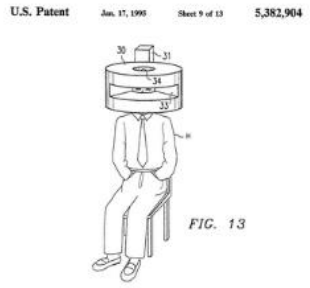


Dipolar electromagnet design

A variety of other related but interesting configurations exist. The FONAR 360°™ (below left) is no longer in production, but was a dipolar resistive electromagnet whose pole pieces protruded from the ceiling and the floor. There was no visible magnetic yoke as it was embedded the walls of the room,

thus allowing 360° access to the patient. Fonar still produces an Upright MRI for weight-bearing studies that can be placed in other positions as well.

FONAR 360°™ scanner at Nuffield Orthopaedic Center (Oxford, UK).



A reminder that all MRI scanner designs have not been successful!

Magnetic Field Homogeneity

Homogeneity refers to the uniformity of a magnetic field in the center of a scanner when no patient is present. Magnetic field homogeneity is measured in parts per million (ppm) over a certain *diameter of spherical volume (DSV)*. For example, a 3.0T magnet may guaranteed to have a homogeneity of <1 ppm over a 40 cm DSV. This means that no two points within ± 20 cm of the magnet isocenter differ in magnetic field strength by more than one part in a million, or by no more than 3.0T x (1/1,000,000) or 0.000003T.

When comparing homogeneity specifications from different vendors, it is important to be sure the quoted DSVs are the same. Magnetic homogeneities will always look better if smaller DSVs are quoted, and inhomogeneity increases dramatically as the DSV is increased. For example, a scanner that has a homogeneity of 1 ppm over a 40 cm DSV may only have homogeneity of 3 ppm over a 45 cm DSV.

Shimming

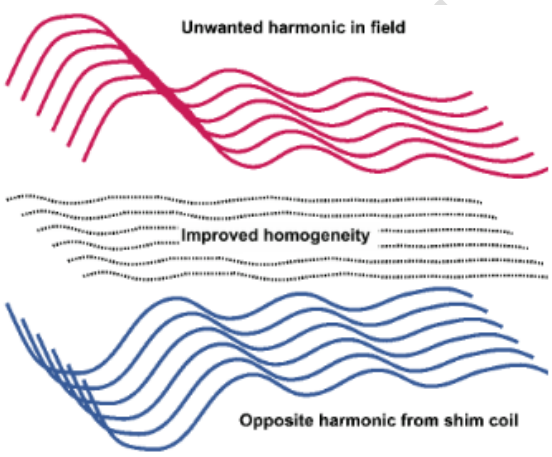
Even following the most rigorous manufacturing tolerances, the homogeneity of an MR magnet arriving fresh from the factory will likely be two orders of magnitude away from ideal specifications levels. Once the magnet is sited in the imaging suite, its field will be further distorted by the presence of metal in pipes, wires, ducts, and structural beams in the immediate environment. Fringe fields of nearby scanners may also affect the field of the newly installed magnet.

Shimming is the process by which the main magnetic field (**Bo**) is made more homogenous. Shimming may be passive, active, or both. In **passive shimming** small pieces of sheet metals or ferromagnetic pellets are affixed at various locations within the scanner bore. In **active shimming**, currents are directed through specialized coils to further improve homogeneity.

In **passive shimming,** small pieces of sheet metal or ferromagnetic pellets are affixed at various locations within the scanner bore to improve homogeneity. Conversely, **active shimming** uses currents directed through specialized coils to generate a "corrective" magnetic field.

Active shim coils can be: 1) ***superconducting***, located within the liquid helium-containing cryostat; or 2) ***resistive***, mounted on the same support structure as the gradient coils within the room-temperature inner walls of the scanner. Both types of active shims require their own power supplies and are controlled by special circuitry. Some scanners use both types.

The theory underlying all active shimming methods is based on spherical harmonic analysis. Field mapping first identifies various unwanted harmonic components in the inhomogeneous field. For each unwanted spherical harmonic component in the uncorrected magnetic field, a carefully controlled supplemental magnetic field is generated by passing current through an active shim gradient. This supplemental shim field has the same spatial distribution, but is equal and opposite to the unwanted component. By super-positioning and merging these two opposite magnetic fields together, a neutralization and cancellation of the magnetic field error (inhomogeneity) is affected.



Theory underlying active shimming. Unwanted harmonics in the inhomogeneous field are cancelled/neutralized by a shim component of equal and opposite polarity.

The relative coil positions can be appreciated in the diagram below, but it may also be useful to list them from outside in:

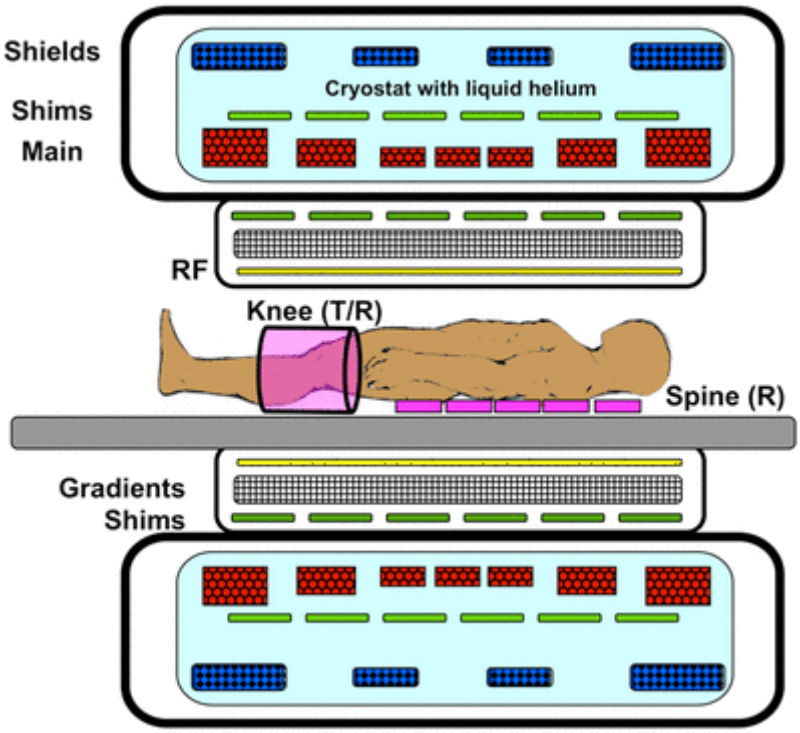
Main field (Bo) Coils (Principal magnet windings plus superconducting shim and shield coils)

→ ***Shim Coils*** (to improve homogeneity)

→ ***Gradient Coils*** (for imaging, including their active shields)

→ ***Radiofrequency (RF)*** Body Coil (transmits B1 field)

→ ***Patient coils*** (primarily to detect MR signal, some are transmit/receive)



Representative cross-section of a superconducting scanner showing nested arrangement of "coils". Both superconducting and resistive shim coils are shown. Two different types of patient coils are also illustrated: a receive only spine coil array, and a transmit/receive knee coil.