Basic Principles of MRI

Nuclear Magnetic Resonance Nuclear magnetic resonance (NMR) is the basis for the imaging technique we now call magnetic resonance imaging (MRI).

The word 'nuclear' had the misleading connotation that nuclear material was used so it was discarded.

Electromagnetic Waves All electromagnetic waves have certain fundamental properties in common:

all emw travel @ speed of light → c

- They all travel at the speed of light ($c = 3 \times 10^8$ meters/second).
- They all have two components - an electric field E and a magnetic field B - that are perpendicular to each other.

$E \perp B$

The electric and magnetic field have the same frequency, both travel at the speed of light, and they are 90° out of phase with each other.

The phase shift is due to the fact that the change in the electric field generates the magnetic field and vice versa - self propagating.

- In terms of vectors, B and E are perpendicular to each other and the propagation factor C is perpendicular to both.

The electrical and magnetic components have the same frequency ω, so we get a vector that is spinning (oscillating) around a point at angular frequency ω.

Remember, the angular frequency, ω, is related to linear frequency, v: $\omega = 2\pi v$

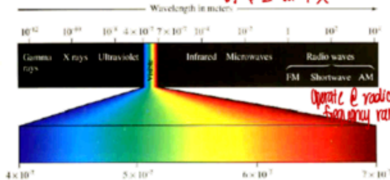
$\omega = 2\pi f$

Basic Principles of MRI

Electromagnetic Spectrum MRI employs much lower energies and frequencies (longer wavelength) than x-ray or visible light.

Electromagnetic Spectrum			
	Frequency	Energy	Wavelength
X-Ray	1.7-3.6 x 10^{18} Hz	30 - 150 keV	80 - 400 pm
Visible Light (Violet)	7.5 x 10^{14} Hz	3.1 eV	400 nm
Visible Light (Red)	4.3 x 10^{14} Hz	1.8 eV	700 nm
MRI	3 - 100 MHz	20 - 200 meV	6 - 60 m

↓f ↓E but ↑λ



operate @ radio frequency range

MRI operates in the radiofrequency range.

The electromagnetic pulses used in MRI to get a signal is called an RF (radio frequency) pulse.

Basic Principles of MRI

atomic nuclei also have spin → means have energy levels related to their spin quantum #

Spin Quantum Number Atomic nuclei (similar to electrons) have specific energy levels related to a property called "spin quantum number", S.

Unpaired protons and neutrons each contribute 1/2 to the overall nuclear spin quantum number, S.

Protons pair with anti-parallel protons canceling out the magnetic properties.
Neutrons do the same.

Unpaired protons and neutrons do not exactly cancel each other's moments.

Nuclei can have more than one unpaired proton and one unpaired neutron (i.e. $S > 1$)

possible: $S > 1$

→ spin quantum #, S → specific energy levels

→ unpaired protons & neutrons contribute ½ to S

→ antiparallel protons cancel out magnetic properties
↓
neutrons too

Basic Principles of MRI

Magnetic Nuclei Some nuclides that have a net magnetic moment are shown below.

Nuclides with a Net Magnetic Moment				
Nuclei	Unpaired Protons	Unpaired Neutrons	S	Magnetic Moment (MHz/T)
^{1}H proton	1	0	1/2	42.58
^{2}H deuteron	1	1	1	6.54
^{13}C	0	1	1/2	10.71
^{14}N	1	1	1	3.08
^{19}F	1	0	1/2	40.08
^{23}Na	1	2	3/2	11.27
^{31}P	1	0	1/2	17.25

determine resonance f to set on on scanner

~ can only see 1 nucleus @ a time

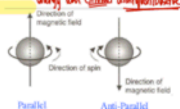
Energy States Atomic nuclei have specific energy levels related to the property called "spin quantum number", S.

@ any T w/o B:
· energy states degenerate, exactly same energy w/ CANNOT distinguish/observe

The number of energy states of a nucleus is determined by the formula:

Number of energy states = 2S + 1

energy states = 2S+1



Example:

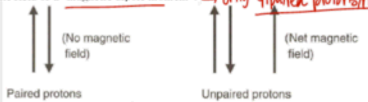
Hydrogen: S = 1/2

Number of energy states = 2(1/2) + 1 = 2

Hydrogen has 2 energy states +1/2 and -1/2

Basic Principles of MRI

Magnetic Dipole Moment: Any nucleus with an unpaired proton or neutron or both has a net magnetic field or a "magnetic dipole moment". ⇒ only unpaired protons/neutrons



Dipole-Dipole Interactions Dipole-dipole interactions refer to interactions between two protons or between a proton and an electron.

MR Imaging of Dipole Moments Any nuclei with an odd number of protons or neutrons can be used for imaging via MR.

The proton, an isotope of hydrogen, is most commonly used for imaging because of its abundance

In MRI, we primarily image the protons in water (H_2O), since ~ 60% of the body is water.

Under certain conditions we can also image the protons in fat $-(CH_2)_n-$

Basic Principles of MRI

Net Magnetization In the absence of an external magnetic field, no magnetization is produced by protons since they are randomly oriented.

each proton still possesses spin vector but summed direction is about 0

PROTONS

No $\vec{B}$ → randomly oriented no magnetization

In the presence of an external magnetic field B_0, the protons act like bar magnets and line themselves up with the magnetic field.

PROTONS

yes $\vec{B}$ → align ∥ (↑, ↓) to $\vec{B}$ field

⇒ equal ~ of ↑ and ↓

Approximately half of the protons point up and half point down. Eventually, slightly more spins point up to make the net magnetization point in the direction of B_0.

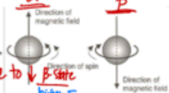
⇒ over time, slightly more in direction of B_0 field (↑)

Proton (Spin) Density Magnetization depends on the tissue being imaged, specifically the density of protons (i.e. protons per unit volume).

It is not the absolute number of protons in the tissue that is important, but rather the number of protons that are rotationally mobile enough to be MR visible. magnetization depends on tissue imaged ⇒ proton density

Basic Principles of MRI

α β



Boltzmann Distribution: There is a slight excess of spins in the ground state (α, spin up, parallel) relative to excited state (β, spin down, anti-parallel).

Boltzmann D: excess of ↑ α-state (lower E) relative to ↓ β-state (higher E)

The number of spins in each state is determined by the Boltzmann Distribution:

$$N_\beta / N_\alpha = \exp -(\Delta E / kT)$$

upper E; lower E; Temp.; Boltz constant

$$\frac{N_\alpha}{N_\beta} = e^{-\frac{\Delta E}{kT}}$$

At room temp, at a field strength of 1.5 T, this corresponds to an excess of about 6 in a million.

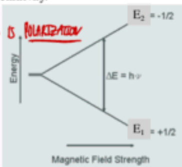
6 α-state (lower) than β-state (higher)

Thus polarization in MR is small and this leads to relatively low sensitivity.

Δ is so small

The degree of polarization is proportional to the energy level difference which in turn is proportional to the strength of the magnetic field and the gyromagnetic ratio.

diff. in states is POLARIZATION



polarization ∝ ΔE ∝ γ

$$\Delta E = h\omega = 2\pi\, h\, \gamma\, B_0$$

$= h\nu = \frac{hc}{\lambda}$

$= h\omega$

Energy Level Difference

Gyromagnetic ratio

Strength of external magnetic field

$\omega = 2\pi f$

γ ∝ magnetic moment → specific to specific nucleus

proton really high γ

Basic Principles of MR

rotational motion

Larmor Equation: The rate at which a proton precesses around the external magnetic field is called the angular precession frequency or the Larmor frequency:

$$\omega_0 = \gamma B_0$$

Angular precessional frequency

Gyromagnetic ratio

Strength of external magnetic field

The gyromagnetic ratio is a proportionality constant that is fixed for any given nuclide. For the proton,

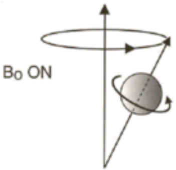
$$\gamma = 42.6\ MHz/Tesla$$

$\omega_0 = \gamma B_0$

Basic Principles of MRI

Precession When a proton is placed in a large magnetic field B_0, it not only rotates about its own axis but also begins to "wobble" or precess about the axis of B_0 as a result of angular momentum.

Remember this
Unpaired and paired



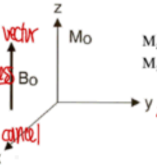
→ precess around B_0, wobble

Basic Principles of MRI

Net Magnetization vector If a sample (patient) is in a magnet, all of the spins are lined up along the axis of the external magnetic field B_0 about which they are precessing (z-axis).

The net magnetization vector, M_0, is the vector sum of all the individual spins and does not precess. This is because the spins are all out of phase and when summed the lateral (x and y) components cancel out.

Σ magnetization vector
→ Does NOT precess
x-and
y- components cancel



$M_z = M_0$
$M_x = M_y = 0$

Σ Mx and Σ My cancel
out due to processing protons

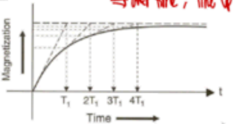
Basic Principles of MRI

Magnetization vs. Time Immediately after being placed in a magnetic field, the spins are randomly oriented.

⇒ Initially, spins are randomly oriented

Over time, the spins line up in parallel or antiparallel to the magnetic field creating a net magnetization.

⇒ over time, line up ↑ or ↓ creating Σ magnetization



At $t = T_1$, the net magnetization is at 63% of its maximum

At $t = 2T_1$, the net magnetization is at 86% of its maximum

T_1 Relaxation Time The time constant describing the growth of magnetization, M, when protons are exposed to a magnetic field is called the T_1 relaxation time and is described by the equation:

$$\Sigma M_0 = M_z = 1 - e^{-t/T1}$$

T_1 = 63% lined up
$2T_1$ = 86%
$4T_1$ = 95%

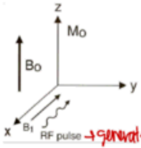
The time constant of this curve (T_1) depends on:

- The type of tissue being imaged
- The strength of the magnet (As the strength of the magnetic field B_0 decreases, then T_1 of the tissue also decreases).

$\omega_0 = \gamma B_0$

Basic Principles of MRI – Effects of a Pulse

Radio Frequency (RF) Pulse: If we transmit an RF pulse along the x-axis perpendicular to the net magnetization vector M_0 (i.e. B_0), then the protons witness a new magnetic field B_1.



B_0 = a very strong external magnetic field (e.g. 1.5 T)

B_1 = a very weak magnetic field generated by the RF pulse (e.g. 50 mT)

→ generates B_1
$\omega_0 = \gamma B_0$
→ at resonance f of Larmor equation

Resonance The RF pulse is in the form of an oscillating magnetic field. If the frequency ω_2 of the RF pulse matches the frequency of proton precession about B_0 then resonance occurs.

The frequency of the RF pulse must match the proton precessional frequency ω_0 in order for the RF pulse to have any effect on the protons at all (i.e. $\omega_2 = \omega_0$ = Larmor frequency).

Because the magnetic field strength of B_1 is much weaker than B_0, the precession frequency ω_1 of the spins around B_1 is much slower than the precession frequency around B_0.

Basic Principles of MRI – Phase Coherence

Before the RF pulse, the protons precess about the z-axis but they are out of phase and thus have no net transverse direction

before RF, precess about z-axis, NO transverse direction

After the RF pulse, the protons are introduced to a new magnetic field B_1 and consequently they will also tend to line up with the new magnetic field and will then be in phase.

line up w/ B_1 (transverse)

y pulse → x M_0

−x pulse → y M_0

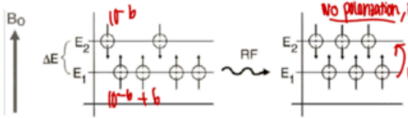
The RF pulse creates a transverse magnetization (M_{xy}) as more and more protons line up (phase coherence increases). The net result is a loss of longitudinal magnetization (i.e. in the direction of B_0).

↑ duration / intensity of Rf, ↓ longitudinal

Basic Principles of MRI – Quantum description

Longitudinal Magnetization Prior to an RF pulse, the protons that are aligned with the external magnetic field are in one of two energy states.

Those in the lower energy state (E_1) are lined up with (i.e. parallel to) the magnetic field B_0, and those in the higher energy state (E_2) are aligned in the opposite direction.



10^{-6}

$10^{-6}+6$

No polarization, No longitudinal

As energy is added by the RF pulse, some of the protons from the lower energy state are boosted to the higher energy state. This happens only on resonance or at the Larmor frequency.

only @ resonance / larmor f

Basic Principles of MRI- Nutation

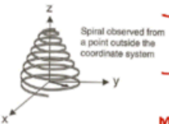
Following an RF pulse, there is a net increase in transverse magnetization (M_{xy}) as more an more protons line up (phase coherence increases).

The spins now experience the effects of 2 magnetic field (B_0 and B_1) and thus simultaneously precess at 2 different frequencies, ω_0 and ω_1.

Recall that: $\omega_1 << \omega_0$

and $B_1 << B_0$

This results in a spiral motion of the net magnetization vector from the z-axis into the x-y plane. This spiral motion is called "nutation".



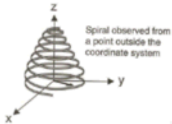
→ experience two $\vec{B}$ (B_0 & B_1)

→ and two ω precessing (ω_0 & ω_1)

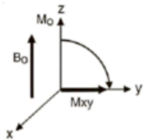
move from z → xy plane in spiral motion (nutation)

Basic Principles of MRI

Rotating Frame of Reference For someone outside the coordinate system (following an RF pulse) it would appear that the net magnetization vector is rapidly precessing around the z-axis as it slowly spirals down into the x-y plane.



However, it the observer is located within a rotating coordinate system that is moving at the same frequency as the protons precessing around the z-axis, then it appears that the net magnetization vector follows a simple arc as it falls from the z-axis into the x-y plane.



rotating frame of reference

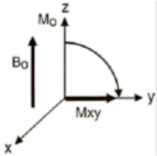
→ rotates of ω_1

→ f of B_1 pulse

Basic Principles of MRI

90° RF Pulse The net magnetization in the direction of B_0, prior to an RF pulse, is called M_0.

When an RF pulse is applied the magnetization vector flips 90° into the x-y plane. The component of M_0 in the x-y plane is called M_{xy}.



90° RF moves $M_0 \to M_{xy}$

If the entire vector flips into the x-y plane, the the magnitude of x-y equals the magnitude of the vector M_0. This is called a 90° flip.

$|M_{xy}| = |M_0|$ @ 90°

The pulse that causes the 90° flip is called the 90° RF pulse.

Remember: There is only a flip into the x-y plane when RF equals the Larmor frequency.

RF = larmor f

Basic Principles of MRI

Flip Angle The angular frequency at which protons rotate 90° about the x-axis is given by the Larmor equation:

$$\omega_1 = \gamma B_1$$

Angular precessional frequency about the x-axis

Gyromagnetic ratio

Magnetic field associated with the RF pulse

Since B_1 is much weaker than B_0, the precession frequency ω_1 around the x-axis is much slower than the precessional frequency ω_0 around B_0.

The flip angle of the magnetization vector into the the xy-plane can be calculated as:

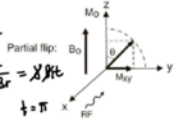
γB_1

$$\theta = \omega_1 \tau = \gamma B_1 t$$

$90° = \omega_1 \tau = \omega_1 \left(\frac{\pi/2}{\gamma B_1}\right) = \frac{\omega_1 \pi}{2\gamma B_1} = \gamma B_1 t$

$180° = \omega_1 \tau = \frac{\omega_1 \pi}{\gamma B_1} = \gamma B_1 t$

$t = \pi$



The time necessary to flip the protons 90° ($\pi/2$) into the x-y plane at a given RF strength B_1 can be obtained with the equation

$$\tau_{\pi/2} = \pi/2 \,/\, \gamma B_1$$

Basic Principles of MRI

inverts

↓↓ 180°

180° RF Pulse A 180° pulse has twice the power or duration of a 90° pulse. After a 180° pulse, the longitudinal magnetization vector is inverted (-z-direction).

After a 180° pulse, the excess north-pointing spins are boosted from the low energy state to the high energy state.

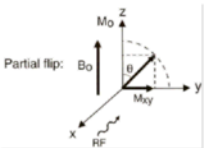
↑ (low state) to ↓ (high state)

A 180° pulse exactly reverses the equilibrium northward-pointing excess without inducing phase coherence (i.e. transverse magnetization).

No transverse

Partial Flip In the case of a partial flip (< 90°), the component of magnetization ending up in the x-y plane (i.e. M_{xy}) can be written as:

$$M_y = M_0 \cdot \sin\theta$$



perturbed → relax

Basic Principles of MRI

Relaxation Times The term "relaxation" means that the spins are relaxing back into their lowest energy state and will get out of phase with each other.

These event result from two simultaneous but independent processes occurring after the RF pulse is turned off:

T_1 Relaxation Time T_1 relaxation time (also called longitudinal relaxation or the spin-lattice time) is the time it takes for the spins to realign along the longitudinal (z) axis.

T_1: the M_z component (M_0) slowly recovers along the z-axis.

1-2 s

T_2 Relaxation Time T_2 relaxation time (also called transverse relaxation or the spin-spin relaxation time) is the time it takes for the spins to dephase in the x-y plane.

T_2: The M_{xy} component of the magnetization vector dephases rapidly.

ms

dephase
→ spins randomly and seep into xy plane rather than phase coherence

Basic Principles of MRI – T_1 Relaxation

T_1 Relaxation Time: T_1 relaxation time (also called longitudinal relaxation or the spin-lattice time) is the time it takes for the spins to realign along the longitudinal (z) axis.

T_1 relaxation time is also called spin-lattice relaxation time because it refers to the time it takes for the spins to give the energy they obtained from the RF pulse back to the surrounding lattice.

The M_z component grows at a rate characterized by T_1: $M_z(t) = M_0\left(1 - e^{-t/T_1}\right)$

regrowth

So apply $B_0 \rightarrow M_0$

apply RF $\rightarrow B_1$

turn off RF $\rightarrow$ relaxes back

(We've seen this before!)

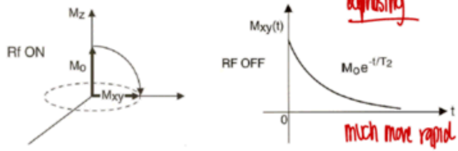
Basic Principles of MRI – T_2 Relaxation

T_2 Relaxation Time The rate at which the transverse vector M_{xy} decays is characterized by the time constant T_2 where

$$M_{xy}(t) = M_0 e^{-t/T2}$$

T_2 relaxation is often called the transverse or spin-spin relaxation time.

dephasing



much more rapid

Important: The recovery of magnetization along the z-axis and the decay of magnetization within the x-y plane are two independent processes occurring at two different rates.

T_2 decay occurs 5 to 10 times more rapidly than T_1 recovery.

Basic Principles of MRI

Dephasing: Immediately following the 90° pulse all spins are in phase; they are all lined up in the same direction.

There are two phenomena that will make the spins get out of phase:

- Interactions between spins – T_2
- External field inhomogeneities – T_2^* due to magnet

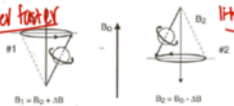
Spins go out of phase because they all begin to precess at slightly different frequencies

Interactions Between Individual Spins When two spins are next to each other, the magnetic field of one proton affects the proton next to it.

For example, assume one proton is aligned with the field (B_0) and the other is against it.

little faster

little slower



Proton #1 is exposed to the magnetic field B_0 plus a small magnetic field created by the other proton (ΔB)

It will process at a frequency slightly faster than the Larmor frequency (ω_0+).

Proton #2 is exposed to the magnetic field B_0 minus a small magnetic field created by the other proton (ΔB)

It will precess at a frequency slightly slower than the Larmor frequency (ω_0-).

This interaction is an inherent property of every tissue and is called a spin-spin interaction. The effect of spin-spin interactions depends on the proximity of the spins to each other.

gases T_2 longer in gases

Basic Principles of MRI

T_2 versus T_2^* It should be emphasized that T_2 and T_2^* are distinct.

T_2 decay depends primarily on:

1. Spin-spin interactions

T_2 also depends on diffusion (i.e. how rapidly spins spread out and leave the lattice); however, this is a minor factor in comparison to spin-spin interactions.

T_2 of a tissue, because it depends only on spin-spin interactions, is fixed - we have no control over what the spins do to each other.

T_2^* decay depends on both:

1. External magnetic field
2. Spin-spin interactions

T_2^* is not fixed – it depends on the homogeneity of the external magnetic field.

T_2^* is always less than T_2 (i.e. T_2^* decay is always faster than T_2 decay), but they are related

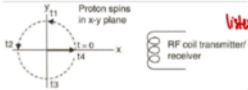
$$\frac{1}{T_2^*} = \underbrace{\frac{1}{T_2}}_{\text{Relaxation rate of Tissue}} + \underbrace{\gamma \Delta B}_{\text{Magnetic field inhomogeneity}}$$

If there was no inhomogeneity (i.e. $\gamma\Delta B = 0$), then $T_2 = T_2^*$

External Magnetic Field Inhomogeneity External magnetic field inhomogeneity always exists to some extent and makes protons in different locations precess at different frequencies because each spin is exposed to a slightly different magnetic field strength.

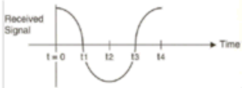
Basic Principles of MRI - Detection

RF Coil An RF coil is an electrical device generally composed of multiple loops of wire that can either generate (transmit) a magnetic field or detect a changing (oscillating) magnetic field as an electric current induced in the wire.



listen to spins as they produce current through RF coil

After a 90° pulse, the magnetization vector rotates in the x-y plane at frequency ω_0.



received frequency → resonance at ω_0

When the magnetic field is in the same direction as the RF coil receiver, a very large signal is induced in the RF receiver coil. When the magnetic field is perpendicular to the direction as the RF coil receiver, there is no signal induced.

A graph of the received signal will look like a sinusoidal curve with a frequency ω_0.

Received Signal Intensity The signal received by the RF coil is an induced current.

The signal is dependent on the number of mobile protons in the tissue, regardless of TR and T_1.

→ signal intensity depends on # of mobile protons

Basic Principles of MRI - FID

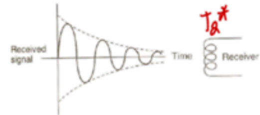
Free Induction Decay (FID) After a 90° pulse, there is a spiral-like decay of transverse magnetization because of spin dephasing.

Decay of magnetization in the transverse (x-y) plane

The change in magnetization along the z-axis does not induce a signal.

→ Only measure magnetization in xy, z has no effect

The signal picked up by the receiver is a decaying oscillating signal.



T_2^*

NOTE: the time constant describing the decay is T_2^* not T_2

This signal is called a free induction decay (FID) and is described mathematically as

$$M_{xy}(t) = M_0 e^{-t/T_2^*} (\cos \omega_0 t)$$

larmor f

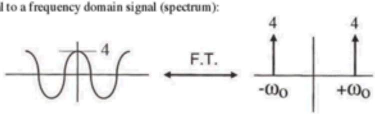
Transverse magnetization

Exponential decay

Oscillating component

Basic Principles of MRI

Fourier Transformation: The FID is Fourier transformed to convert from a time domain signal to a frequency domain signal (spectrum):



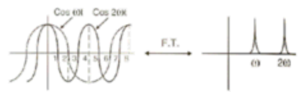
Each oscillating signal in the FID gives rise to a pair of lines displaced by $\pm\omega_0$ from the central frequency. The signal intensity is proportional to the strength of the signal in the FID.

The time domain and frequency domain signals exist as Fourier pairs:

$$G(\omega) = \int_{-\infty}^{\infty} g(t)\, e^{-i\omega t}\, dt \qquad G(t) = 1/2\pi \int_{-\infty}^{\infty} g(\omega)\, e^{i\omega t}\, d\omega$$

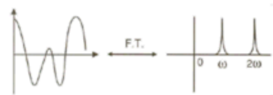
Basic Principles of MRI

The Fourier transform can be used to provide the frequency domain of very complex time domain signals.



Only the positive frequencies are shown.

If we sum these two curves.



Complex – hard to interpret

Discrete frequencies – easier to interpret

Basic Principles of MRI - Summary

- Wait some time for spins to reach equilibrium in the magnetic field (TR $\geq 3 * T_1$):

$$M_z(t) = M_0\left(1 - e^{-t/T_1}\right)$$

- Apply a pulse or series of pulses to bring magnetization into the x-y plane:

$$\theta = \omega_1 \tau = \gamma B_1 t$$

- Wait for spins to dephase (TE):

$$M_{xy}(t) = M_0 e^{-t/T2}$$

- Turn on the receiver and observe the free induction decay (FID):

$$M_{xy}(t) = M_0 e^{-t/T_2^*}(\cos\omega_0 t)$$

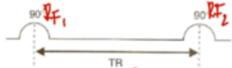
-Apply processing filters and Fourier transform FID to observe spectrum/image:

$$G(\omega) = \int_{-\infty}^{\infty} g(t)\, e^{-i\omega t}\, dt$$

Define: TR - repetition time; TE - echo time

Basic Principles of MRI – Pulse Sequences

Repetition Time (TR): A pulse sequence is a pulse or series of pulses, applied sequentially, followed by signal acquisition. The time interval between applications is called TR (the repetition time).

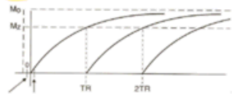


- Before a 90° pulse ($t = 0^-$), the magnetization vector (M_0) points along the z-axis.
- Immediately after a 90° pulse, the magnetization vector $M_{x,y}$ lies in the x-y plane, with no component along the z-axis. $M_{x,y}$ has magnitude M_0.
- With time, magnetization recovers along the z-axis (due to T_1) and is lost in the xy-plane (due to T_2).
- At time TR, we apply another 90° pulse and the existing longitudinal magnetization vector (M_z) flips back into the x-y plane. This process can be repeated.
- The magnitude of the magnetization vector M_z at the time TR can be calculated

$$M_z(t) = M_0\left(1 - e^{-t/T_1}\right)$$

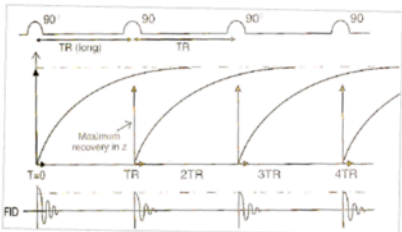
$$\downarrow$$

$$M_z(t) = M_0\left(1 - e^{-TR/T_1}\right)$$



Basic Principles of MRI – Saturation Recovery

Saturation Recovery Pulse Sequence This pulse sequence involves trying to recover all the longitudinal magnetization before we apply another 90° RF pulse. TR is long: > 3 x T_1.

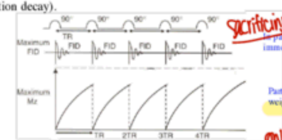


The FID is measured immediately after each 90° pulse, .

Basic Principles of MRI – Partial Saturation

Partial Saturation Pulse Sequence Application of a 90° pulse, waiting for a short TR, then applying another 90° pulse.

Measurements are obtained immediately after the 90° RF pulse. The signal received is an FID (free induction decay).



sacrificing some signal

In partial saturation, the signal is measured immediately after the 90° pulse.

Partial saturation is used to produce a T_1-weighted image.

only magnetic dephasing

At time t = 0, longitudinal magnetization is flipped into the x-y plane.

At time t = n TR, the longitudinal magnetization has not recovered to equilibrium before it is flipped into the x-y plane again.

NOTE: There may be no residual transverse magnetization M_{xy} at time nTR just before the next pulse. T_2 is several times shorter than T_1. M_{xy} has fully decayed.

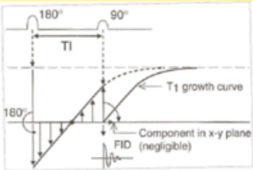
dephase decay shorter than recovery M_z

Basic Principles of MRI – Multiple Pulse Sequences

Inversion Recovery Pulse Sequence This is the simplest multiple pulse sequence. It involves first applying a 180° RF pulse, waiting a period of time, TI, before a 90° pulse is applied. The FID is collected and the pulse sequence is repeated after TR.

TI = Interval Time.

After the 180° pulse, the magnetization vector points in the -z-direction. Over time, the magnetization vector gets smaller in –z, goes through zero and grows in the +z-direction.



allows to collect Ti map

After a time TI a 90° pulse is applied, flipping the longitudinal magnetization vector into the x-y plane. The magnitude of M_{xy} depends on the amount of longitudinal magnetization that has recovered during time TI.

Basic Principles of MRI – Inversion Recovery

Recovery Curves in Inversion Recovery When using the inversion recovery pulse sequence, there are two different exponentially growing recovery curves.

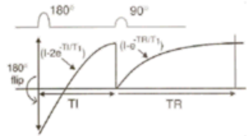
1. Recovery after the 180° RF pulse.
2. Recovery after the 90° RF pulse.

- The T_1 recovery curve following the 180° RF pulse starts at $-M_0$ and grows exponentially

$$M_z = M_0\left(1 - 2e^{-TI/T_1}\right)$$

- The T_1 recovery curve following the 90° RF pulse starts at 0 and grows exponentially

$$M_z = M_0\left(1 - 2e^{-TI/T_1}\right)\left(1 - e^{-TR/T_1}\right)$$

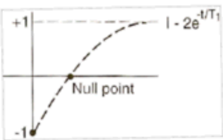


Basic Principles of MRI – Inversion Recovery

Null Point Following a 180° pulse, the point at which the signal crosses the zero line is called the null point.

how to calculate T1

At the null point the signal intensity is zero. The time is denoted TI(null). We can calculate the null point as follows:



$$Signal\ Intensity = 0 = 1 - 2e^{-TI(null)/T_1}$$

$$\ln 1 = \ln\left(2e^{-TI(null)/T_1}\right)$$

$$0 = \ln 2 + \ln\left(e^{-TI(null)/T_1}\right)$$

$$\ln 2 = \frac{TI(null)}{T_1}$$

$$TI(null) = 0.693T_1$$

$$T_I(null) \cong 0.70T_1$$

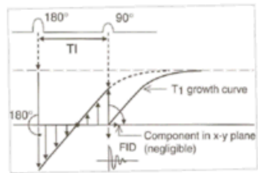
The null point can be used to estimate the T_1 relaxation time!

Basic Principles of MRI – Measuring T_1

Measurement of T_1 is performed using an inversion recovery pulse sequence.

1. Wait TR = 5 * T_1, until the magnetization is fully relaxed to equilibrium
2. Use a 180° pulse to invert magnetization
3. vary TI systematically
4. Apply 90° pulse to convert recovering magnetization into x-y plane
5. Magnetization after 90° pulse is given by: $M_{xy} = M_0\left(1 - 2e^{-TI/T_1}\right)$

6. A plot of ln M_{xy} vs TI yields a straight line with slope $1/T_1$

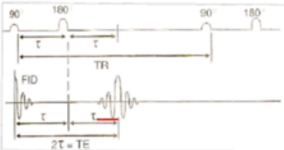


Basic Principles of MRI – the Spin Echo

Spin-Echo Pulse Sequence The spin echo is the most commonly used pulse sequence in MRI. It is used to eliminate the dephasing caused by external magnetic field inhomogeneities.

As a result of a 90° RF pulse, the magnetization vector M_z is flipped into the x-y plane. Spins begin to dephase with time constant T_2^*.

Application of a 180° pulse at time TE/2 leads to a reversal in the direction of the spins, causing the spins to rephase and refocus in the form of a spin echo at time TE.



τ is the time between the two pulses.

TE is the time between the first pulse and the echo.

TE = 2τ .

TE (Echo Delay Time) is defined as the time between the initial pulse and the peak of the echo in a multi pulse or spin echo pulse sequence.

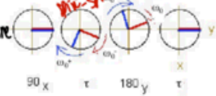
Basic Principles of MRI – the Spin Echo

Spin-Echo Pulse Sequence

At a certain time τ after a 90° pulse, when the spins have gotten out of phase, a 180° pulse is applied.

All the spins flip 180° in the x-y plane, and they continue precessing, but now in the opposite direction.

• slower
• faster
starting
dephasing
spin-spin interactions

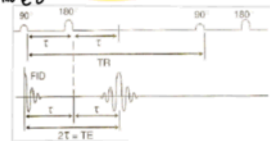


τ is the time from the 90° pulse to the 180° pulse.

τ is also the time from the 180° pulse to the point of maximum rephasing (i.e. the echo)

2τ is the echo delay time TE. The time after the 90° pulse when we get maximum signal again.

After an equal time τ, the spins will be completely in phase again. This is why the 180° pulse is often called the rephasing or refocusing pulse.

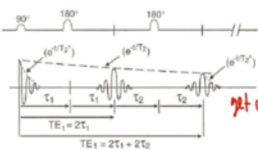


Basic Principles of MRI – Multi Spin Echoes

Multi Spin-Echo Pulse Sequence

In a spin echo sequence, after the first echo, the spins will begin to dephase again.

A second 180° pulse applied at time τ_2 after the first echo will allow the spins to rephase again at time $2\tau_2$ or TE_2.



The time from the 90° pulse to the first echo is TE_1.

The time from the 90° pulse to the second echo is TE_2.

Each individual echo decays with time constant T_2^*.

get decay of echo

The echo train decays with time constant T_2.

Ideally, we would like to regain all of the signal from the original FID. However, in practice we only regain the signal lost due to fixed external magnetic field inhomogeneities by applying a refocusing 180° pulse.

↑ SNR ↓ aquisition time

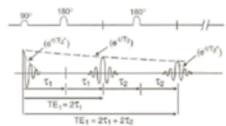
Dephasing caused by spin-spin interaction i.e. by T_2 cannot be regained since these interactions are not fixed (i.e. they fluctuate randomly).

FID decays by T_2* star
returns so only decays due T_2

Basic Principles of MRI – Measuring T_2

The Multi Spin-Echo Pulse Sequence is used to measure T_2

to measure T_2



If we join the points of maximum signal at the peak of the echoes are we will get an exponentially decaying curve with a time constant given by T_2.

In other words, the decay of the curve describing the echo train is given by

$$S = S_0 \exp(-TE/T_2).$$

$S_0 \propto M_0$

A plot of ln (S) vs TE yields a linear plot with slope = $1/T_2$.

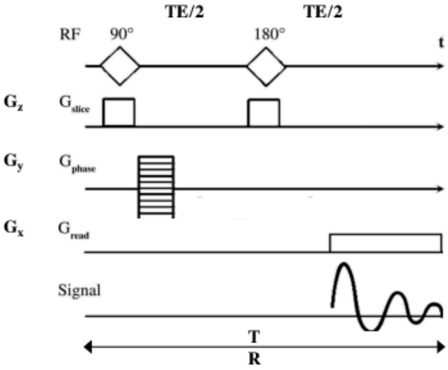
NOTE: The decay or each individual FID is given by e^{-T2^*}, which is more rapid than T_2 decay due to the influence of magnetic field inhomogeneities.

MRI - Image Construction

A sequence timing diagram is a graphical notation of the pulses and gradients applied during an MR pulse sequence.

One line for RF and for gradients in 3-orthogonal directions, one line for detected signal.

Gradient A gradient is an applied magnetic field that changes from point to point - usually in a linear fashion.



Spin echo is the most basic and commonly used pulse sequence for MRI

.

Contrast is generated for different tissues by altering TE and TR.

Three types of commonly applied contrast: T_1, T_2 and proton density.

Sequence timing diagram for a standard spin echo pulse sequence.