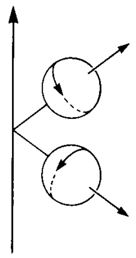
Department of Medical Physics & Biomedical Engineering

MPHY0001: Introduction to Medical Imaging
Magnetic Resonance Imaging

2. Magnetic resonance imaging

The story so far:

- Every Hydrogen nucleus effectively contains a tiny bar magnet (magnetic moment). In an external **B** field, rules of quantum mechanics demand that these adopt either a "spin up" or "spin down" state.



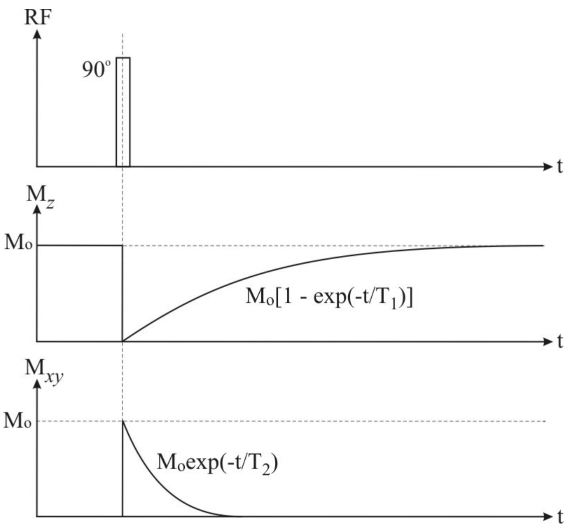
Nuclear alignment in a magnetic field

- Because the moment ("spin") cannot align exactly with **B**, each moment experiences a torque which causes it to precess at the Larmor frequency $\omega = \gamma B$.
- The total field **M** due to all H nuclei aligns parallel with **B**, although the individual spins cannot.
- **M** can be realigned away from **B** direction by adding another $\mathbf{B_1}$ field which rotates at exactly the Larmor frequency.

The story so far:

- A $\mathbf{B_1}$ field which rotates **M** into the *xy* plane is known as a 90° RF pulse.
- Immediately following 90° pulse:
  a) Equal number of spins are in "spin up" and "spin down" states.
  b) All spins are precessing "in phase",
- **M** then spirals back to original alignment parallel to $\mathbf{B_0}$, due to two "relaxation mechanisms".
- *Spin-lattice relaxation*: higher energy "spin down" states switch back to lower-energy "spin up states" until "thermal equilibrium" is reached (timescale characterised by T_1).
- *Spin-spin relaxation*: spins no longer precess in phase (timescale characterised by T_2).
- Movement of **M** induces a current in a nearby coil.

The story so far:



To make images using the NMR signal, the magnetic field must be different at different locations within the patient.

This is achieved using **magnetic field gradients**.

The idea to use field gradients was independently proposed by Peter Mansfield (Nottingham UK) and Paul Lauterbur (USA), who were jointly awarded 2003 Nobel Prize for Medicine.

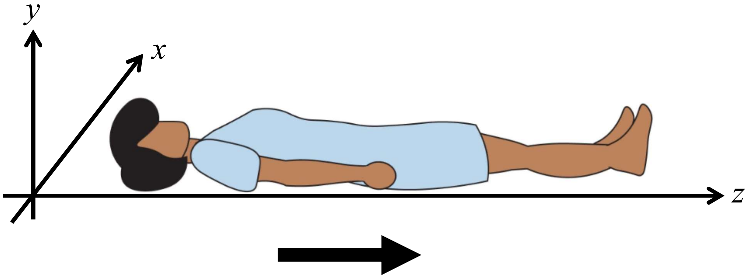
Peter Mansfield Paul Lauterbur

2.1 Magnetic field gradients

To spatially localise signals, magnetic field gradients are used so that main field can vary linearly (and independently) along *x*, *y*, and *z*.

Gradients are defined as:

$$G_z = \frac{\partial B_z}{\partial z} \qquad G_x = \frac{\partial B_z}{\partial x} \qquad G_y = \frac{\partial B_z}{\partial y}$$



$$B(x,y,z) = B_0 + x.G_x + y.G_y + z.G_z$$

2.1 Magnetic field gradients

To spatially localise signals, magnetic field gradients are used so that main field can vary linearly (and independently) along *x*, *y*, and *z*.

Gradients are defined as:

$$G_z = \frac{\partial B_z}{\partial z} \qquad G_x = \frac{\partial B_z}{\partial x} \qquad G_y = \frac{\partial B_z}{\partial y}$$

Thus for gradient along z direction:

$$B = B_0 + z.G_z$$

Thus: $\omega(z) = \gamma\,(B_0 + z.G_z)$

For all three gradients: $\omega(x,y,z) = \gamma\,(B_0 + x.G_x + y.G_y + z.G_z)$

Generating an image involves three components:

slice selection, *phase-encoding*, and *frequency-encoding*.

2.2 Slice selection

An arbitrary plane in three-dimensional Cartesian space is defined by the equation:

$$ax + by + cz = \text{constant.}$$

$$\omega(x,y,z) = \gamma\,(B_0 + x.G_x + y.G_y + z.G_z)$$

$\Rightarrow$ for set of gradients, a plane exists at which B and therefore ω, is constant.

If a 90º pulse ($\mathbf{B_1}$ field) is applied *at that frequency*, **M** for all protons in that plane is rotated into transverse plane. This is known as *slice selection*.

For many imaging applications, selected slice is perpendicular to main field (z) axis, in which case $G_x = G_y = 0$.

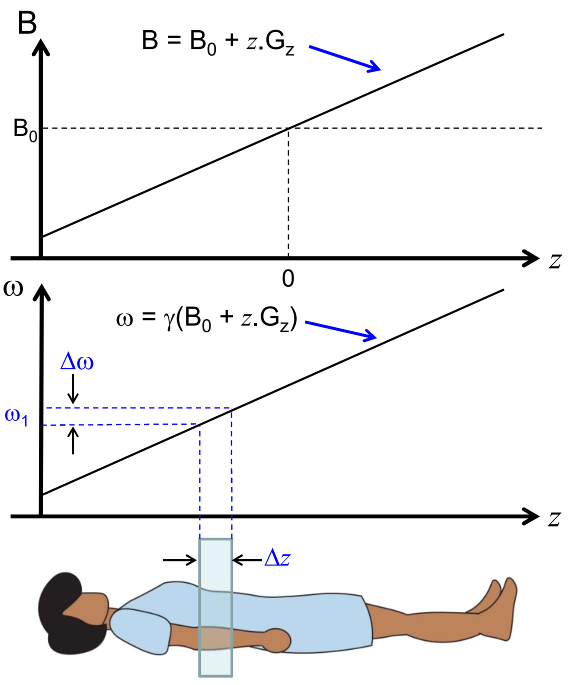
In practice, $\mathbf{B_1}$ field will oscillate over range of frequencies ($\Delta\omega$) known as the *bandwidth*, which depends on pulse length τ:

$$\Delta\omega = \frac{2\pi}{\tau}$$

Selected slice has finite thickness given by:

$$\boxed{slice\,thickness \qquad \Delta z = \frac{\Delta\omega}{\gamma\, G_z}}$$

Different slices can be selected by either changing the gradient or varying the central frequency of RF pulse.

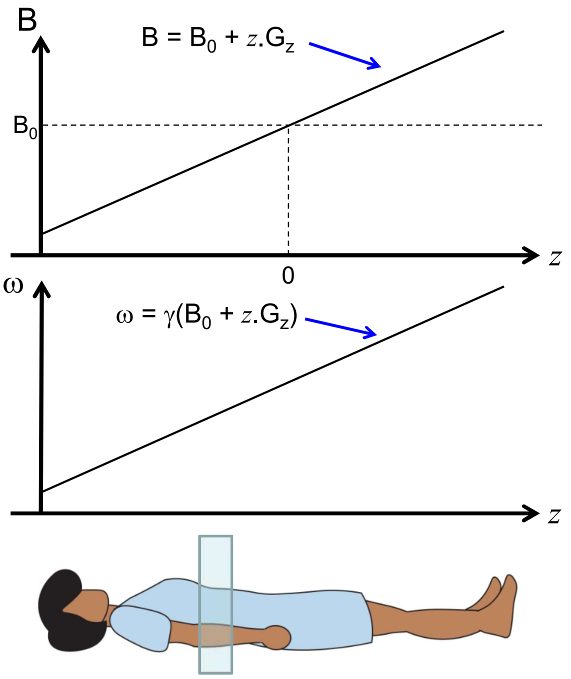


Field B along z axis due to main magnet B_0 and applied gradient G_z.

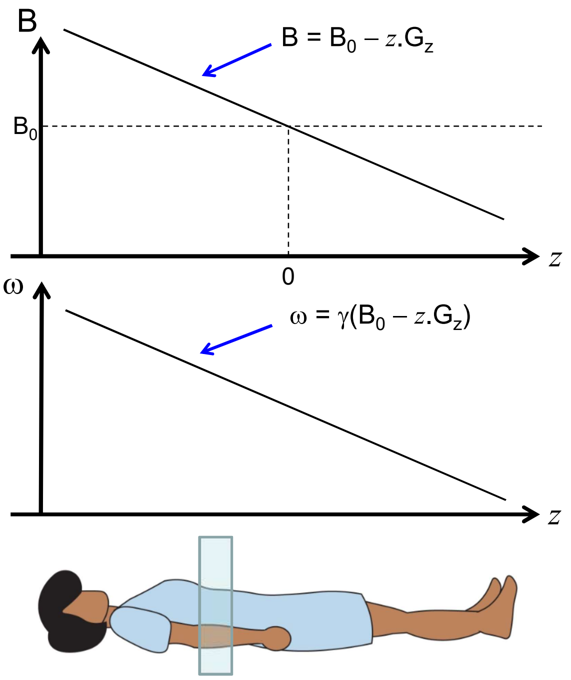
Larmor frequency along z axis: $\omega(z) = \gamma\,(B_0 + z.G_z)$

RF pulse (B_1 field) applied at frequency ω_1 with bandwidth $\Delta\omega$.

$$slice\,thickness \qquad \Delta z = \frac{\Delta\omega}{\gamma\, G_z}$$

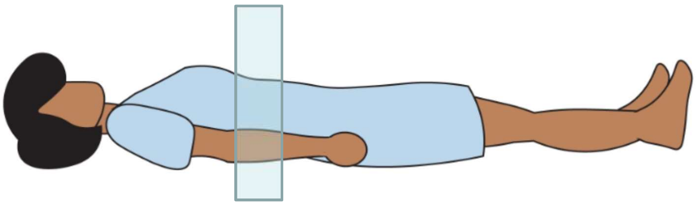


After pulse, protons on one face of slice will be precessing slightly faster than protons on other face.



After pulse, protons on one face of slice will be precessing slightly faster than protons on other face.

To make them rotate in phase again, a *negative gradient* is briefly applied along the same direction.



After slice selection, frequencies and phases of Larmor precessions within slice can be manipulated by application of further gradient fields, so that each volume element ("voxel") within slice is uniquely identifyable.

2.3 Phase encoding

Now consider what happens within the slice after slice selection gradient (e.g. G_z) is switched off, and a perpendicular gradient (e.g. G_y) is switched on for a short period of time Δt.

If we ignore relaxation processes, **M** of protons in selected slice will precess at rate dependent on field strength along y axis:

$$\omega(y) = \gamma\,(B_0 + y.G_y)$$

After time Δt, **M** will have rotated by angle: $\phi(y) = \gamma\,(B_0 + y.G_y).\Delta t$

Consequently a variable phase shift occurs along y axis. This is known as *phase encoding*.

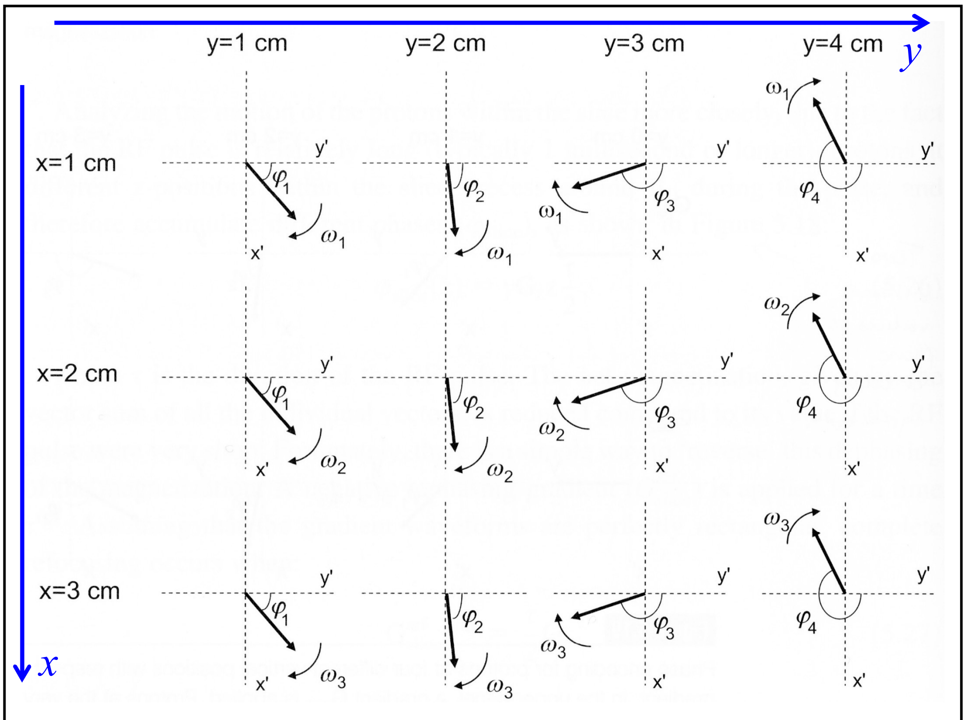
2.4 Frequency encoding

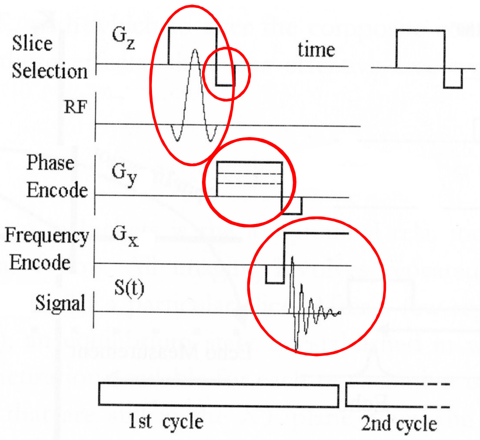
Now consider what happens after phase encoding gradient is switched off, and another perpendicular gradient (e.g. G_x) is switched on.

The magnetisation **M** within the selected slice will now precess at rate dependent on the field strength along the x axis:

$$\omega(x) = \gamma\,(B_0 + x.G_x)$$

Thus, within selected slice, the *frequency* of signal is dependent on position along x-axis, while *phase* is dependent on position along y-axis.





It can be shown that the signal is equivalent to just *one line through 2D Fourier transform* of image.

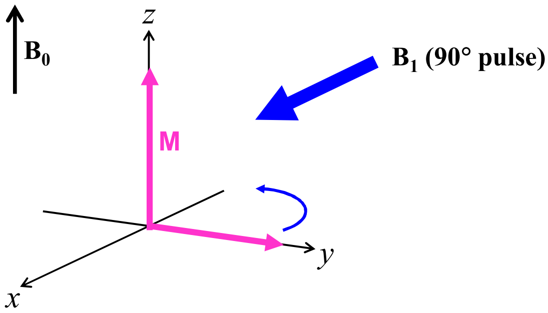
Further lines are acquired by changing phase encoding gradient G_y. Typically 128 or 256 discrete values of G_y are used. Finally, data undergoes an inverse Fourier transform to produce an image.

The contrast of each pixel of image will depend on:

a) Local density of hydrogen nuclei (protons) = $\rho(x,y)$
b) Local value of T_1
c) Local value of T_2 .

The relative contribution depends on timing and ordering of RF pulses and gradient fields. This is known as a “pulse sequence”.

2.5 The spin echo pulse sequence

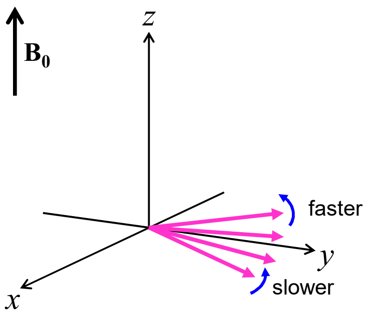


Initially **M** is aligned with z axis.

A 90° pulse flips **M** vector into xy plane.

M will then precess around the z axis.

2.5 The spin echo pulse sequence

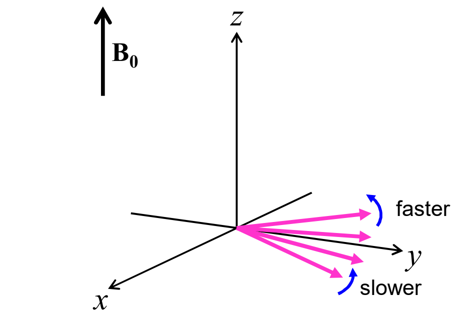


M is the vector sum of many individual spins, which will precess at different rates (loss of phase coherence).

Thus M_{xy} decays over a timescale dependent on T2* .

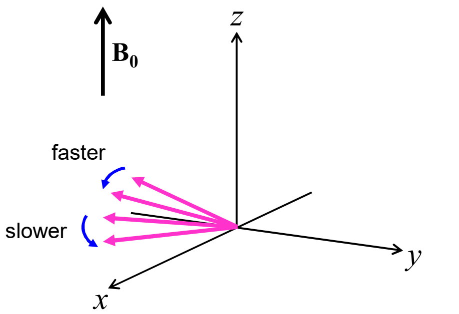
This produces a FID signal which we choose to ignore.

2.5 The spin echo pulse sequence



After a period of time (equal to TE/2) a 180º pulse is applied.

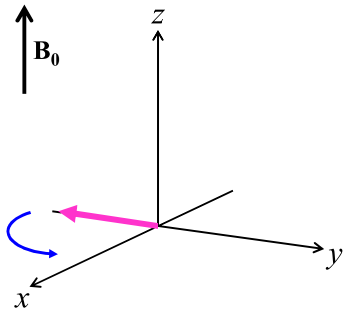
2.5 The spin echo pulse sequence



After a period of time (equal to TE/2) a 180º pulse is applied.

This has the effect of placing the slower spins ahead of the faster spins.

2.5 The spin echo pulse sequence

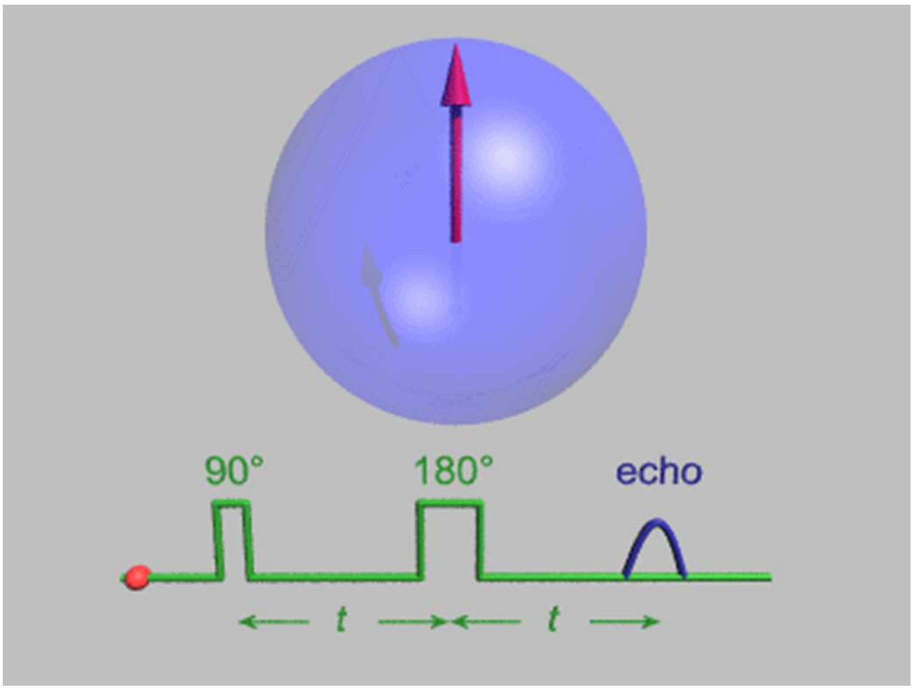


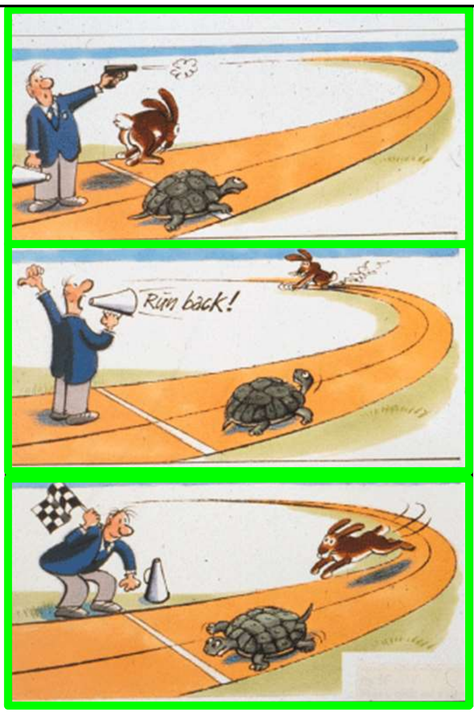
After another period of time equal to TE/2 the faster spins catch up with the slower spins, and all the spins are rotating in phase again.

Thus M_{xy} grows to reach a temporary maximum value.

This creates a signal in a nearby coil, called an "echo".

The echo occurs at a time TE after the 90º pulse.





During formation of echo, effect of field inhomogeneity ΔB is removed. Thus echo magnitude after interval TE is dependent on T2 rather than T2* :

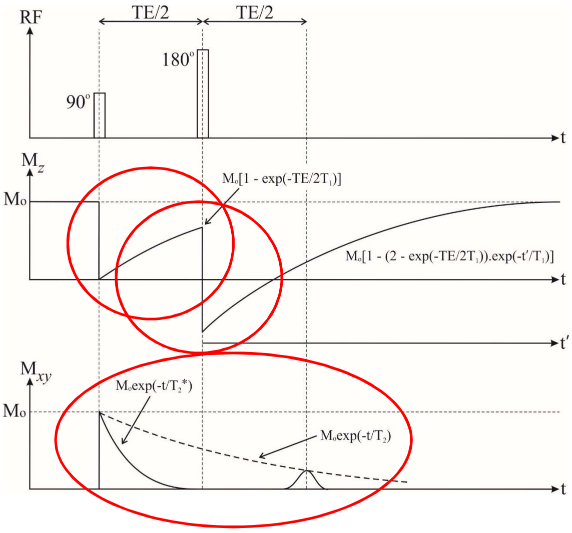
$M_{xy}(TE) = M_0 \exp(-TE/T_2)$.

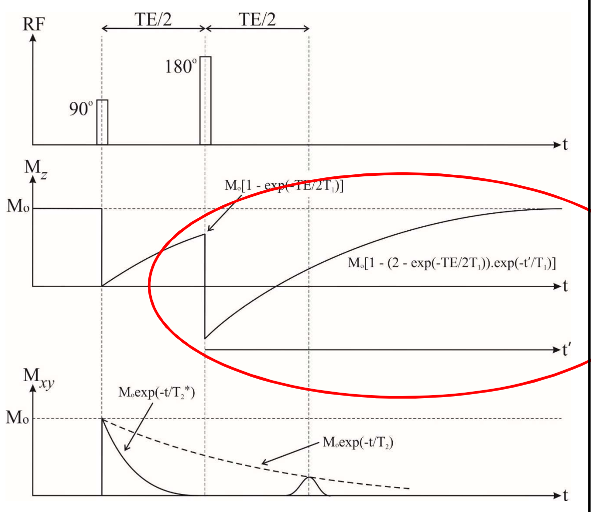
When 180º pulse is applied, M_z will have recovered to:

$M_z = M_0(1 - \exp(-TE/2T_1))$.

The 180º pulse flips M_z :

$M_z = - M_0(1 - \exp(-TE/2T_1))$.





It is shown that at time t' after 180º pulse, the M_z component will be described by:

$$M_z = M_0(1 - \exp(-t'/T_1)) - M_0(1 - \exp(-TE/2T_1)).\exp(-t'/T_1)$$

$$\Rightarrow \quad M_z = M_0\,[1 - (2 - \exp(-TE/2T_1)).\exp(-t'/T_1)]$$

In practice it is necessary to repeat sequence to acquire sufficient signal.

For time between repeats TR, it can be shown that M_z at each 90º pulse is given by:

$$M_z = M_0(1 - \exp(-TR/T_1)).$$

assuming TR >> (TE/2).

Therefore intensity $I(x,y)$ of image using a spin echo sequence is proportional to:

$$\boxed{I(x,y) \propto \rho(x,y).(1 - \exp(-TR/T_1)).\exp(-TE/T_2).}$$

$\boxed{I(x,y) \propto \rho(x,y).(1 - \exp(-TR/T_1)).\exp(-TE/T_2)}$

a) If TR is large: $\exp(-TR/T_1) \approx 0$

$\Rightarrow$ $I(x,y)$ has very little dependence on T_1.

i.e. if M_z has sufficient time to recover back to M_0 it is impossible to determine how quickly that recovery occurred due to T_1.

b) If TE is very short: $\exp(-TE/T_2) \approx 1$

$\Rightarrow$ $I(x,y)$ has very little dependence on T_2.

i.e. a short TE means that there is little time for M_{xy} to decay due to T_2.

c) If TE is very short <u>and</u> TR is very long, T_1 and T_2 dependence virtually disappears, and $I(x,y)$ becomes dependent only on proton density $\rho(x,y)$.

$\boxed{I(x,y) \propto \rho(x,y).(1 - \exp(-TR/T_1)).\exp(-TE/T_2).}$

In summary:

A spin-echo sequence with $TR \approx T_1$ and $TE << T_2$ will produce a "T1-weighted" image.

A spin-echo sequence with $TR >> T_1$ and $TE \approx T_2$ will produce a "T2-weighted" image.

A spin-echo sequence with $TR >> T_1$ and $TE << T_2$ will produce a "proton density" image.

MRI scans of same slice across same brain.

PD: proton density

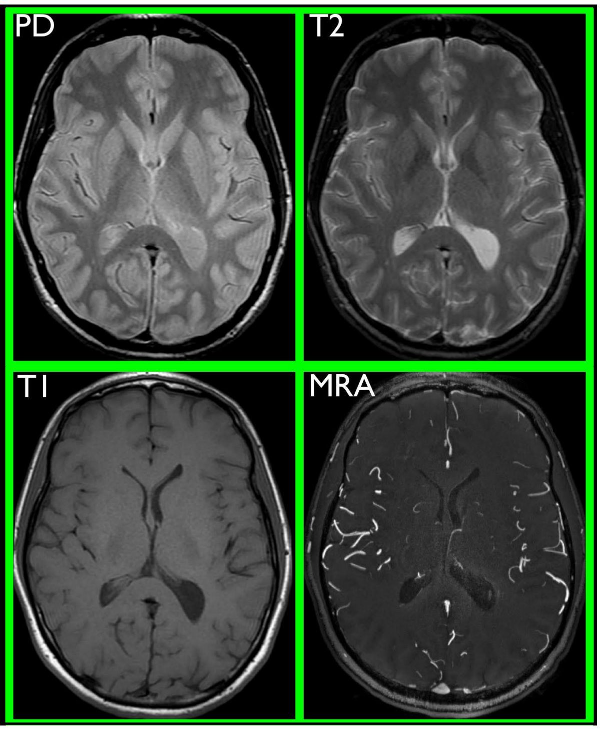
T2: T_2-weighted

T1: T_1-weighted

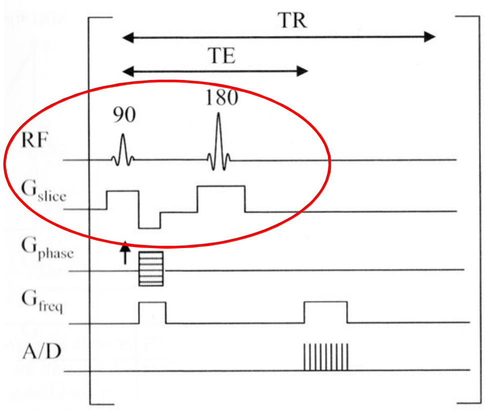
MRA: MR angiography (section 4.2).

Free water has long T_1, so CSF appears dark (low signal) in a T_1-weighted image.

Free water has long T_2, so CSF appears bright (high signal) in a T_2-weighted image.



<u>The spin echo pulse sequence</u>

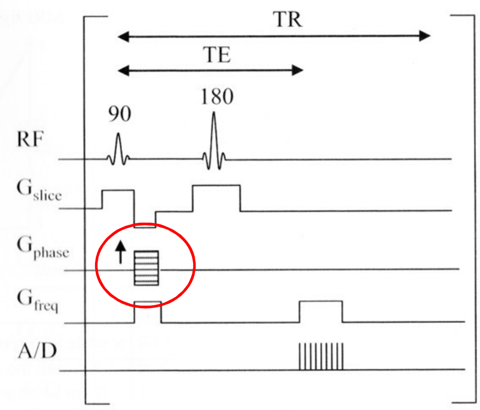


Note that:

The slice selection gradient is applied during <u>both</u> pulses. For 90° pulse, gradient is reversed at end to rephase spins across slice thickness.

For 180° pulse, gradient reversal isn't necessary because the 180° pulse has the same effect.

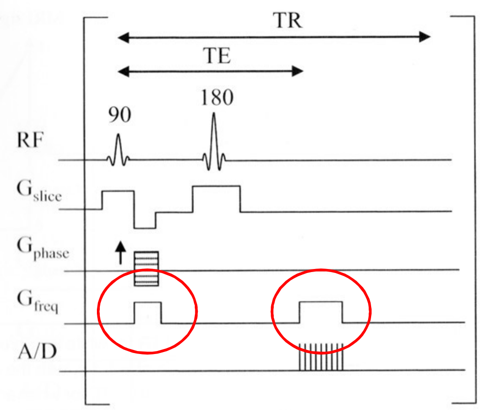
The spin echo pulse sequence



Note that:

The sequence is repeated for different values of phase encoding gradient to acquire a 2D image.

The spin echo pulse sequence



Note that:

Frequency encoding gradient is applied during echo ("readout time").

This same gradient is also applied earlier to ensure that a zero frequency offset occurs exactly in middle of readout time. This is effectively reversed because of subsequent 180º pulse.

The spin echo pulse sequence

This sequence has proven effective for two reasons:

1) echo occurs a long time after RF pulse, giving time to prepare the external coil for detection.

2) T_1 and T_2 weighted contrast can be displayed by selection of TE and TR values.