MRI physics: what are we measuring?

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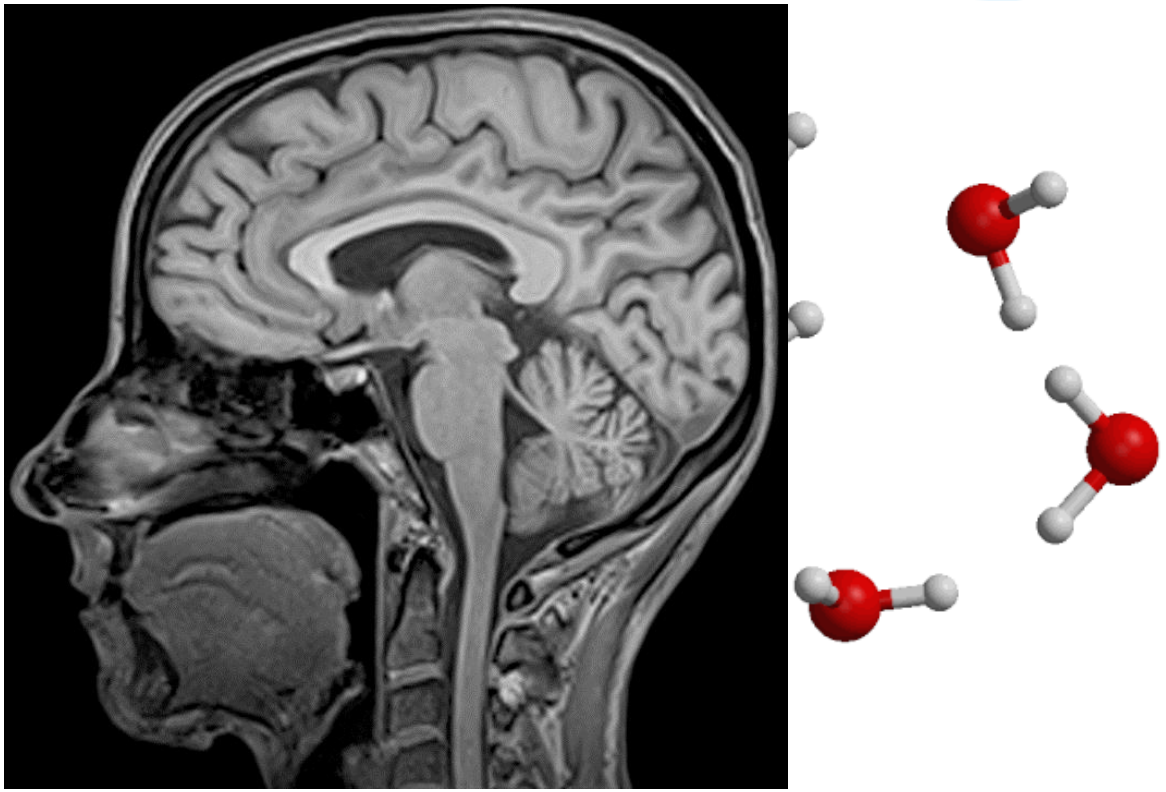
Outline

* Basic Principles of Magnetic Resonance Imaging
* Relaxation and Measurement of NMR Parameters
* Principle of MR Imaging

What is Magnetic Resonance?

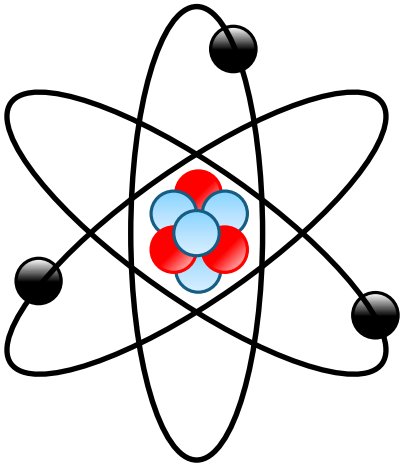
* MRI is based on the principle of Nuclear Magnetic Resonance (NMR).
* Nuclear refers to the fact that MR depends on the properties of the nucleus of an atom.
* NMR refers to study of the resonant behaviour of nuclei under the influence of a magnetic field.

MR Image



What is an atom?

* **Atom** is a basic unit of matter.
* It consists of a nucleus containing **protons** and **neutrons**, surrounded by clouds of negatively charged **electrons**.



Electron
Proton
Neutron

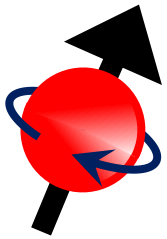
What is spin angular momentum?

* **Nuclear mass** consists of mixture of positively charged protons and electrically neutral neutrons.
* Atomic nuclei with an odd number of protons or neutrons or both have an intrinsic **spin angular momentum** called nuclear spin and exhibit NMR effect, while atomic nuclei with an even number of protons and neutrons have no net spin.

No of Proton	No of Neutron	Spin Angular Momentum
Even	Even	0
Even	Odd	1, 2, 3, …
Odd	Odd	1/2, 3/2, 5/2, …

What is spin angular momentum?

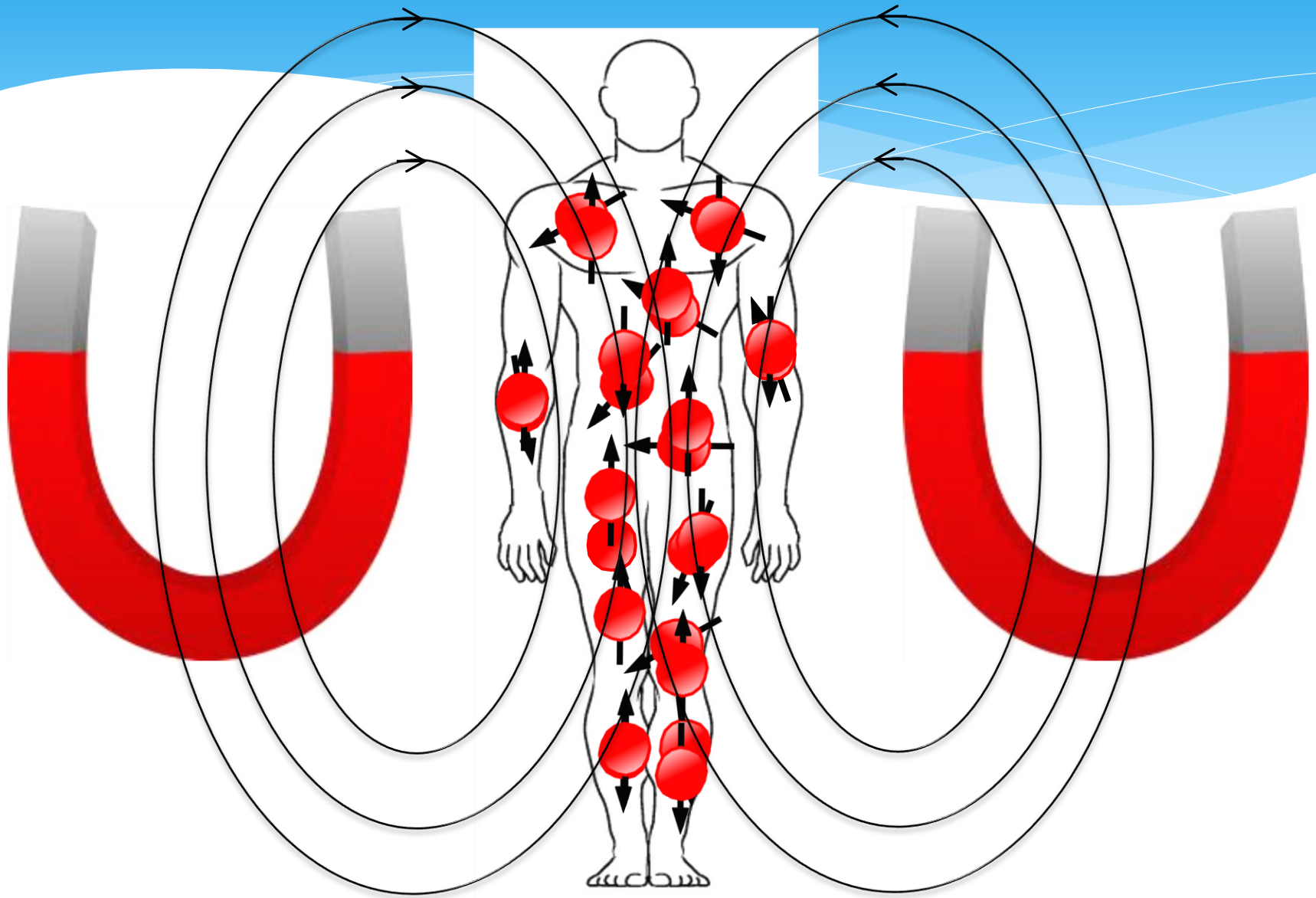
* In order to be NMR sensitive, a nucleus must have spin.
* **Hydrogen** is the most commonly used active nucleus in clinical MRI:
  - ^{1}H is the most abundant isotope in the human body (~75 – 80% of fat and water)
  - Consists of one proton that has a relatively large magnetic dipole moment and high gyromagnetic ratio, which is capable of generating relatively large NMR signals.



Common MRI active nuclei

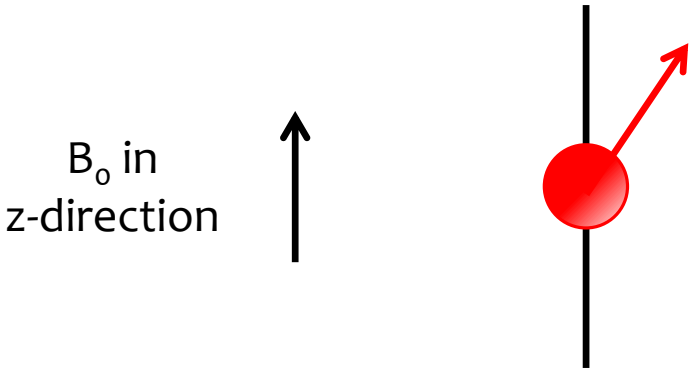
Nucleus	Intrinsic Spins, I	Gyromagnetic Ratio, γ ($MHz\ T^{-1}$)	Natural Abundance (%)
^{1}H	½	42.58	99.98
^{2}H	1	6.54	0.015
^{3}H	½	45.41	0
^{13}C	½	10.71	1.11
^{14}N	1	3.08	99.6
^{15}N	½	-4.32	0.37
^{17}O	5/2	5.77	0.04
^{19}F	½	100	40.05
^{23}Na	3/2	100	11.26
^{31}P	½	100	17.24

NMR Effect



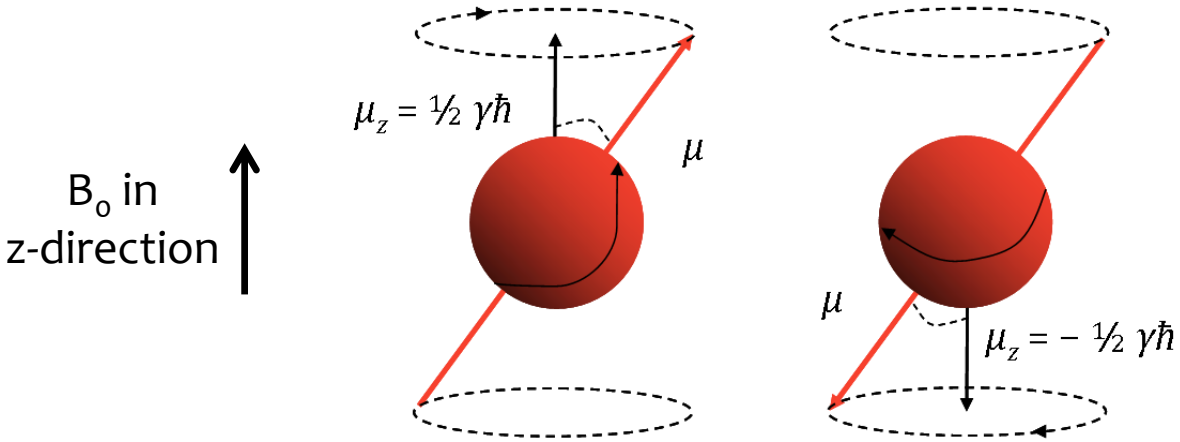
NMR Effect

* The nuclear magnetic moment which arises from spin ½ interacts with the magnetic field, B_o.
* If we align the B_o from bottom to top (z-axis), the proton nuclear magnetic moment will try to align parallel to the z-axis.



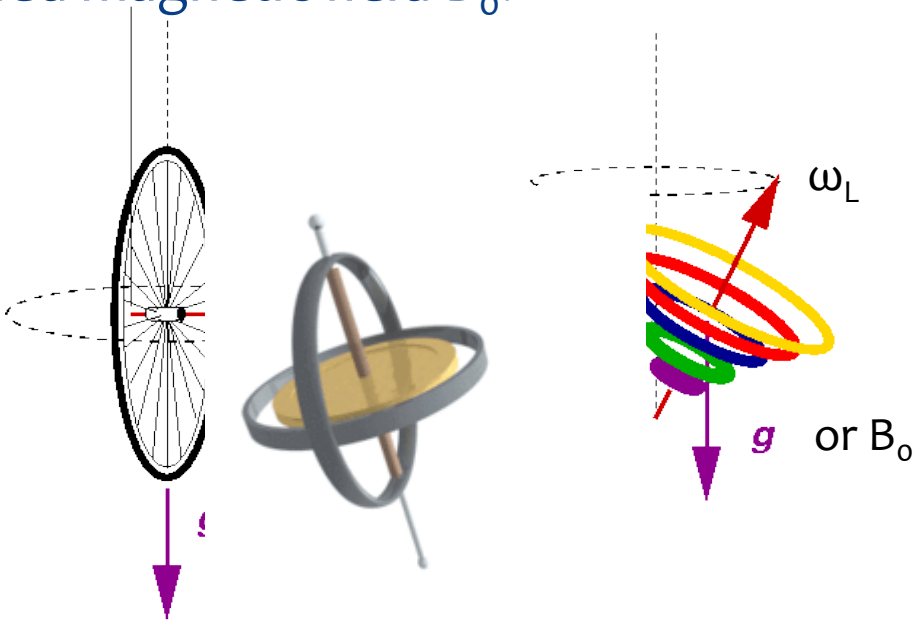
NMR Effect

* On the application of the field, the nuclei will either align parallel or anti-parallel with the magnetic field direction:
  - Parallel = stable low energy state
  - Anti-parallel = unstable high energy state



NMR Effect - Free Precession

* The protons are not static when they align with the field. They experience a torque and precess around the direction of the applied magnetic field B_o.

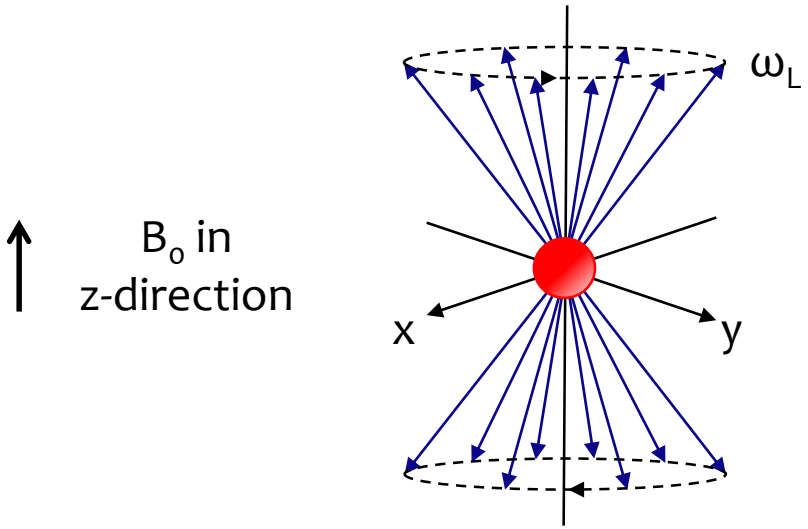


http://www.fis.unipr.it/~derenzi/dispense/pr/wiki.php?n=MRI.SpinPrecession

http://en.wikipedia.org/wiki/NMR_Spin_Precession

NMR Effect - Larmor Frequency

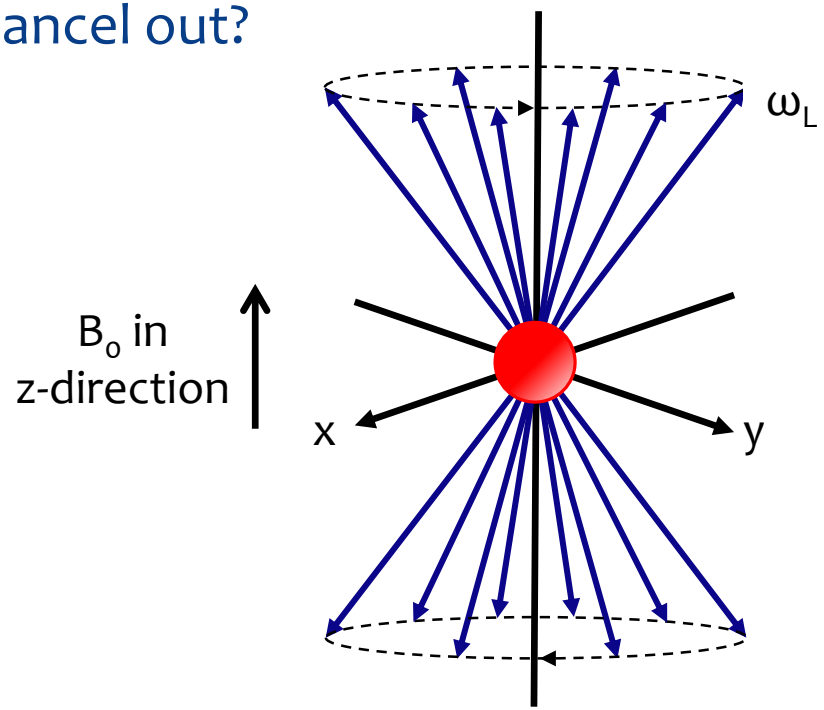
* The angular frequency (ω_L) at which the nuclei precess is called the **Larmor frequency**. It depends on the strength of B_o.
* γ is the gyromagnetic ratio. It is the ratio of the nucleus' magnetic moment to its angular momentum.



$$\omega_L = \gamma B_o$$

NMR Effect

* If we add up all the nuclear magnetic moments in low and high energy state, will the net magnetisation in low and high state cancel out?



Net magnetisation in low energy state

Net magnetisation in high energy state

NMR Effect

* No, the nuclear magnetic moment in the low energy state will be slightly higher than high energy state.

$$\frac{n\uparrow}{n\downarrow} = e^{\frac{\Delta E}{kT}}$$

where n↑ = number of spins in low energy state

n↓ = number of spins in high energy state

Δ E = difference in energy (low and high energy state)

k_B = Boltzmann constant (1.38×10^{-23} JK^{-1})

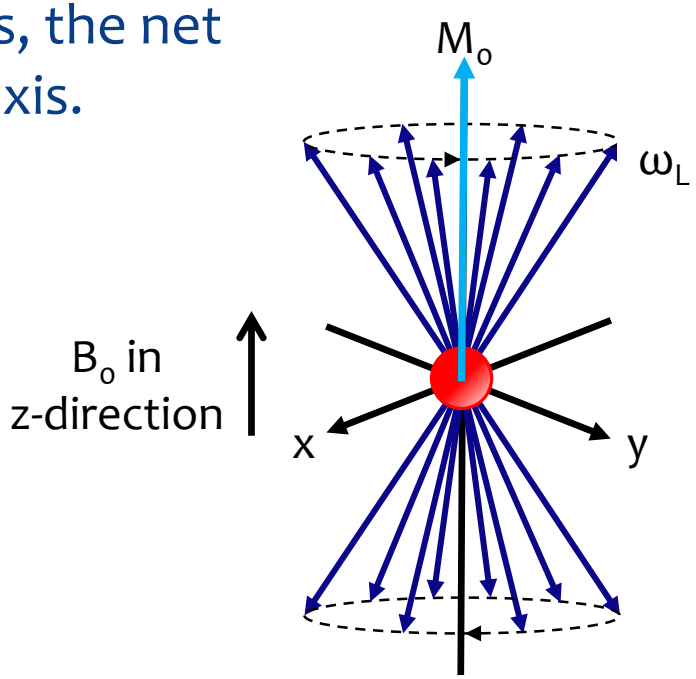
T = absolute temperature

NMR Effect

* The vector sum of the magnetic dipole moments in the resulting spin population difference would give rise to a net magnetisation, $|\mathbf{M}|$ that can be detected for NMR.
* The net magnetisation is low and is the reason for intrinsically low signal to noise ratio in NMR experiments.

NMR Effect

* At equilibrium, many individual spins are distributed around the precessional cone. Hence, there is no net magnetisation along the x- and y-axis, the net magnetisation is aligned with the z-axis.

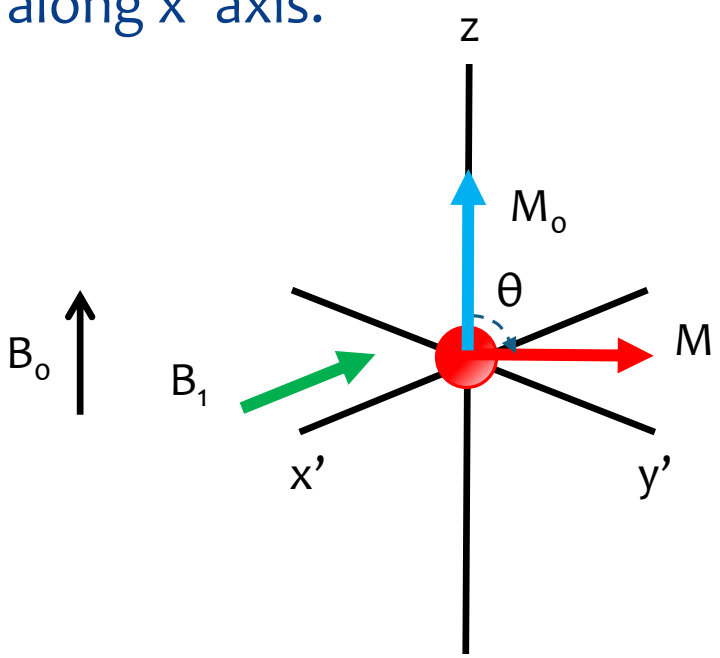


Oscillating Electromagnetic Field

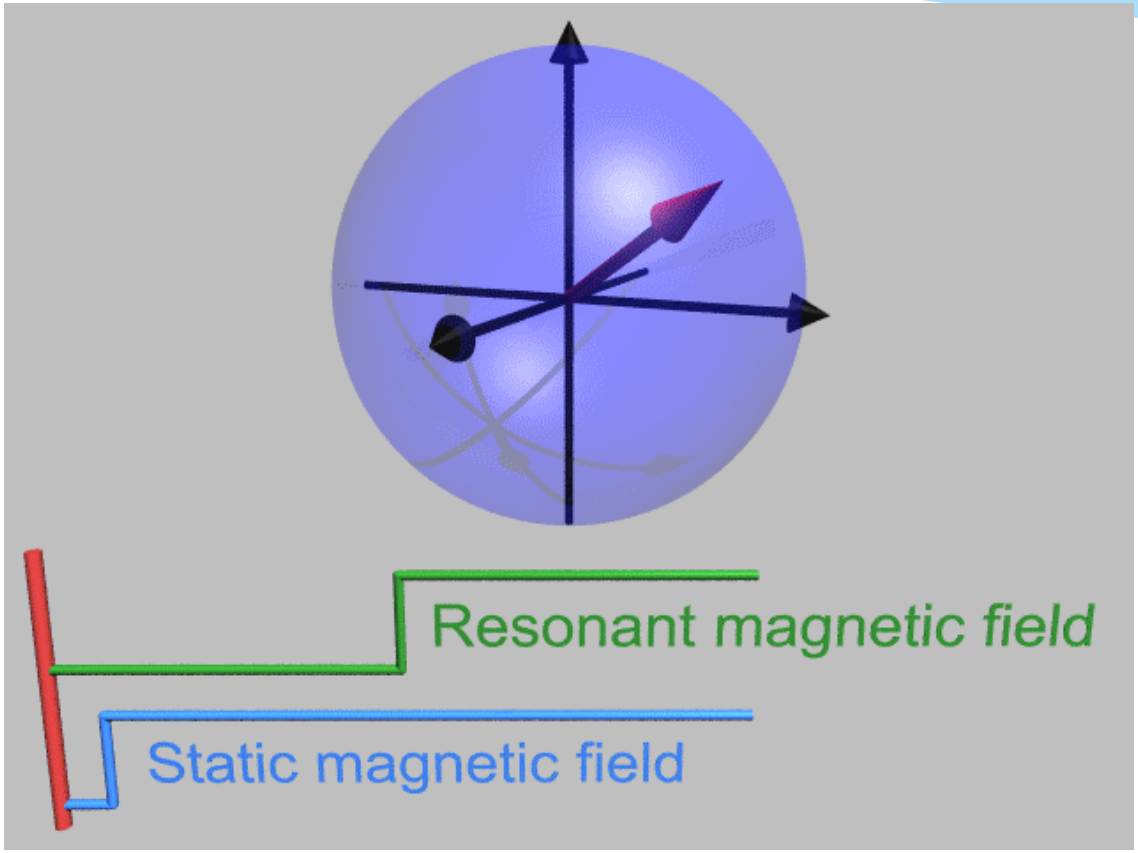
* To detect the NMR signal, the longitudinal magnetisation should be perturbed from its equilibrium state by applying an oscillating radio-frequency (RF) magnetic field, B_1.
* B_1 is perpendicular to B_o and oscillates in phase with the precessing magnetic moments at ω_L.
* The oscillating magnetic field, B_1 is much smaller than the static magnetic field, B_o.

Rotating Frame of Reference

* In contrast to the laboratory frame (x, y, z), the rotating frame rotates clockwise around z-axis in the xy plane (x', y', z) at frequency ω. B_1 is stationary along x' axis.



Rotating Frame of Reference



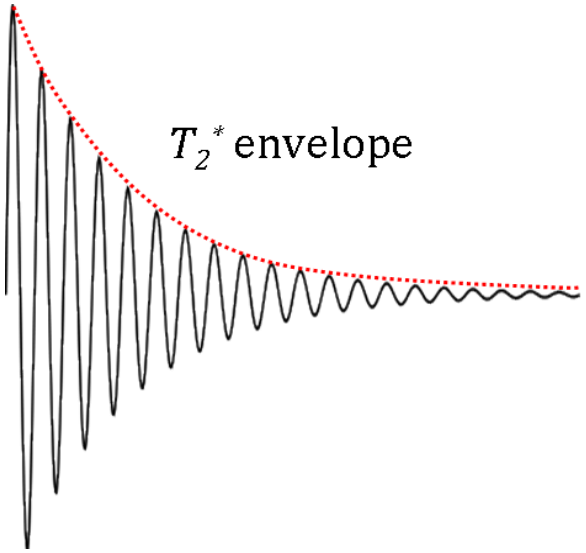
http://commons.wikimedia.org/wiki/File:Animated_Rotating_Frame.gif

Free Induction Decay (FID)

* Application of a 90° pulse or transverse magnetisation causes the spins making up the magnetisation, M to be in phase with each other.
* These spins precess in the transverse plane at ω_L → production of an electrical voltage in a coil tuned to ω_L.
* This voltage is the basis to the observable MRI signal. The signal amplitude depends on the degree to which any longitudinal magnetisation can be tilted to the transverse plane directly after switching off the RF pulse.

Free Induction Decay (FID)

* Hence, this signal will persist until the spins have dephased and lost their transverse consistency. Signal decay is shown here:



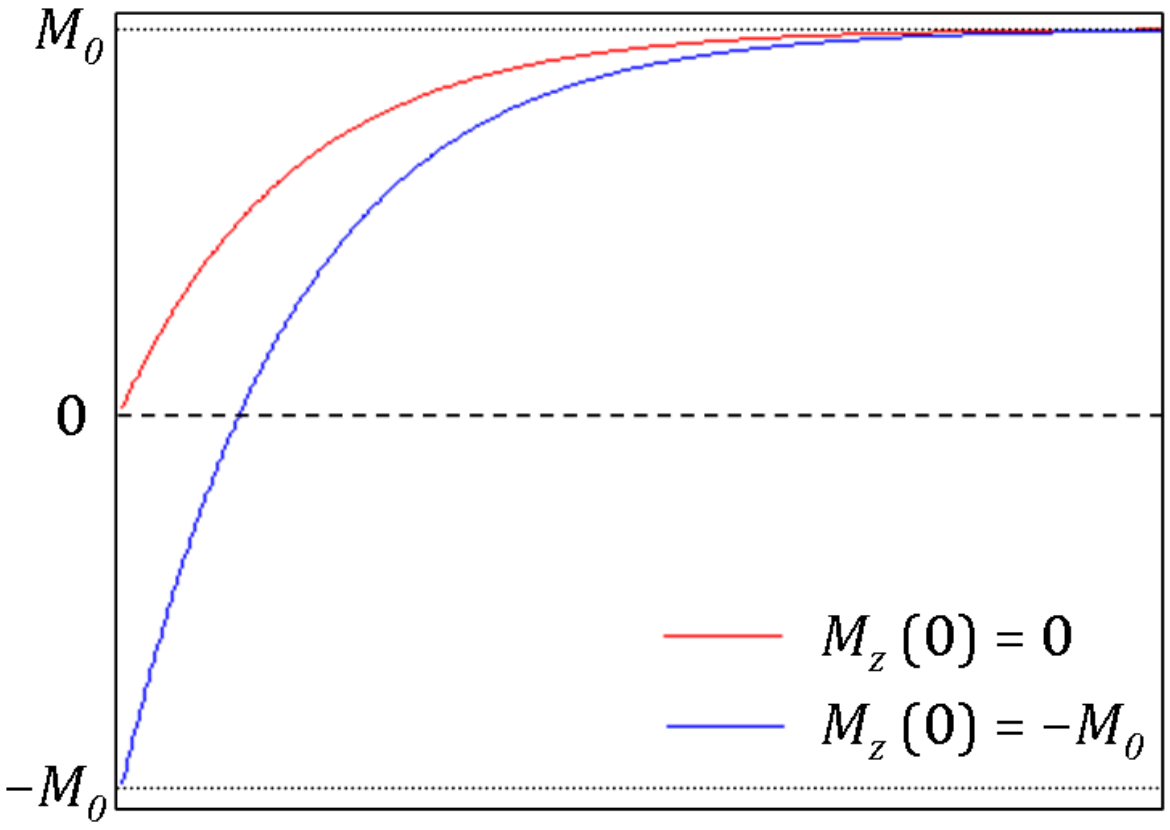
Summary so far… section 1

* Human beings, like the rest of the universe, are made of atoms – a basic unit of matter.
* Atoms are composed of protons, neutrons and electrons. They have an intrinsic property called spin.
* In the presence of an external magnetic field:
  * The spins will be aligned to the field.
  * In order to detect NMR signal, the longitudinal magnetisation should be perturbed from its equilibrium state by applying oscillating RF magnetic field, B_1. $B_1 \perp B_0$ & $B_1 < B_0$.

Longitudinal Recovery (T_1 Relaxation)

* T_1/longitudinal relaxation is the recovery of magnetisation along the z-axis, M_z. It is the time when M_z is ~ 63% M_0.
* This process involves realignment of the magnetic moments with B_0, causing the net magnetisation to align along the z-axis. The rate of the net magnetisation's growth towards its equilibrium value, M_0, is exponential and equivalent to the longitudinal recovery time, T_1.

Longitudinal Recovery (T_1 Relaxation)



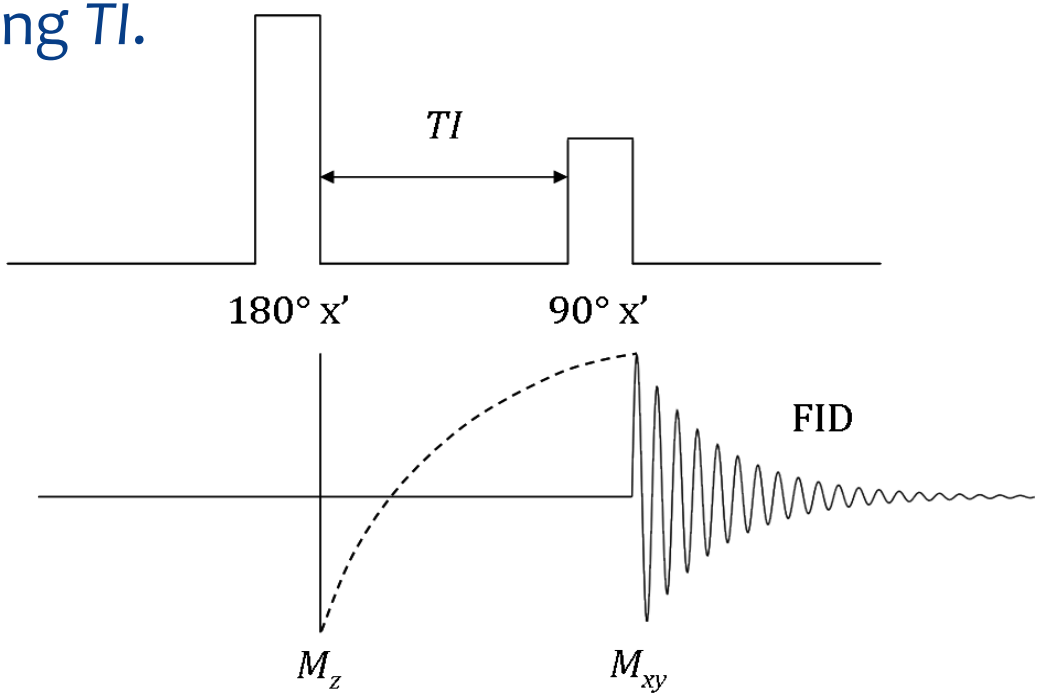
Saturation Recovery

Inversion Recovery

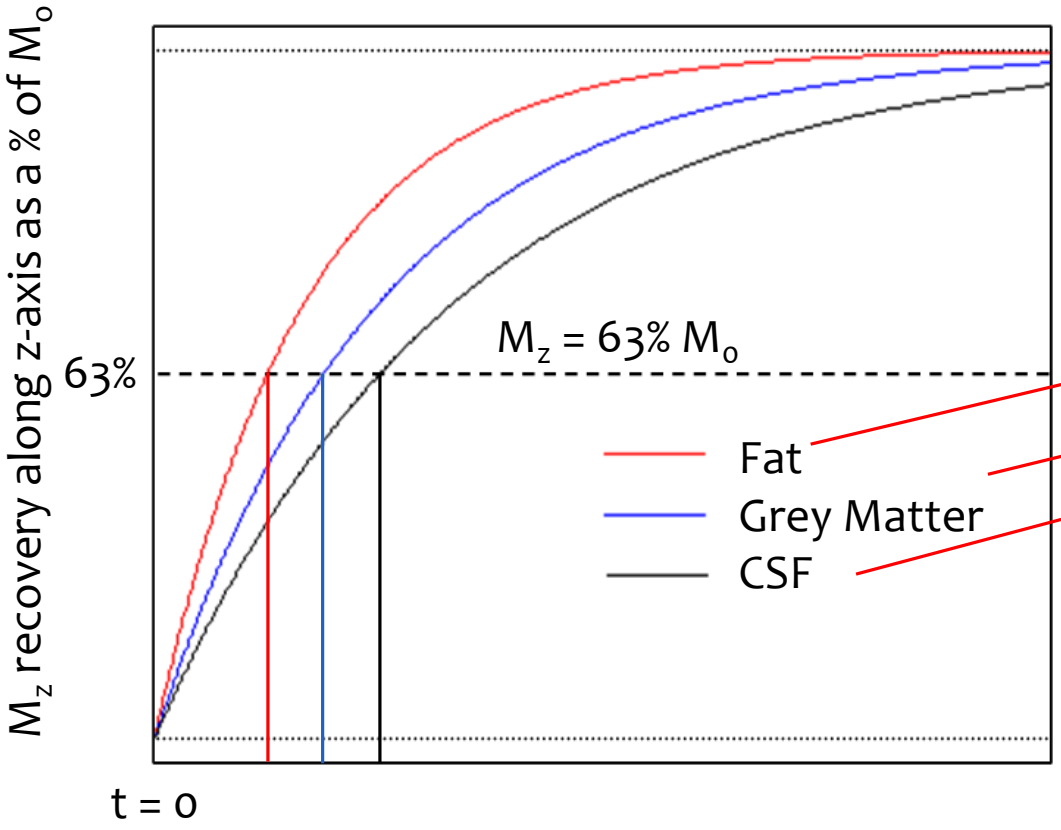
Longitudinal Recovery (T_1 Relaxation)

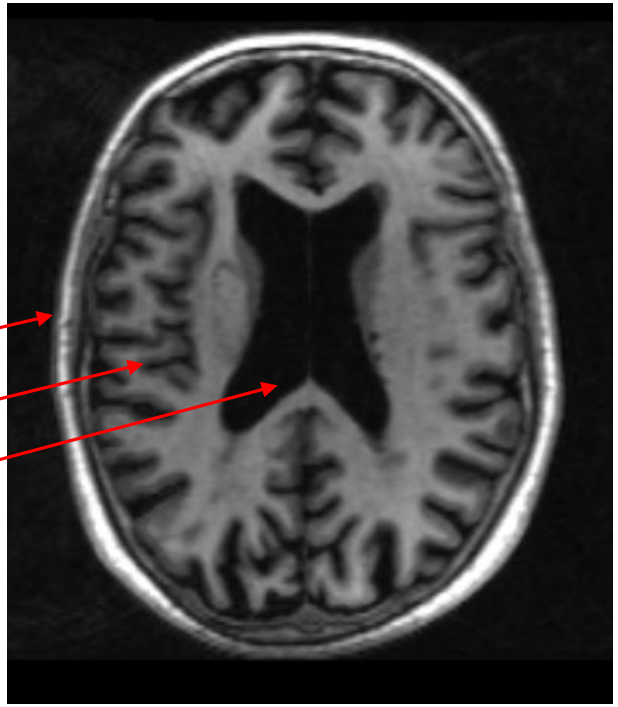
* T_1 can be measured using the inversion recovery sequence, by varying *TI*.

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



Longitudinal Recovery (T_1 Relaxation)

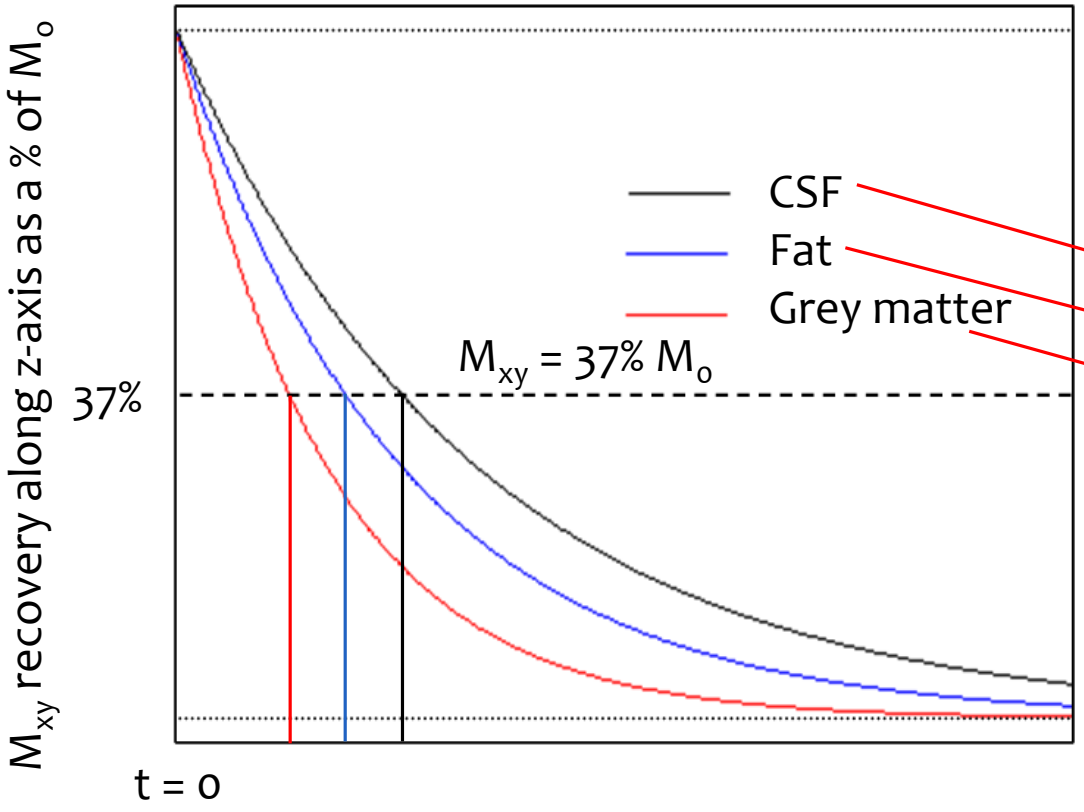


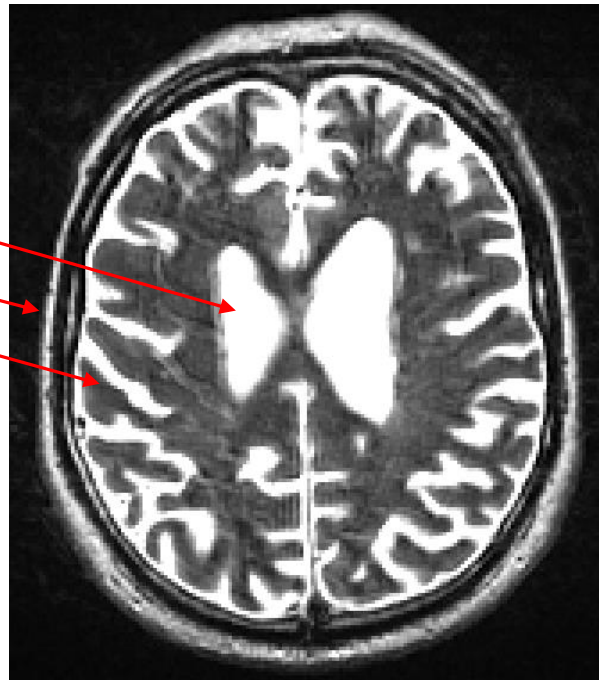


Transverse Decay (T_2 Relaxation)

* Transverse magnetisation/M_{xy}, is usually created following the application of a 90° pulse.
* M_{xy} decays at a rate characterised by a transverse decay time, T_2. It characterises the decay magnetisation in the x'y' plane.
* T_2 is the time required for 37% of the transverse magnetisation to remain for detection.
* It is important to note that the decay of the magnetisation M_y is NOT the inverse of the recovery of M_z along z-axis!

Transverse Decay (T_2 Relaxation)

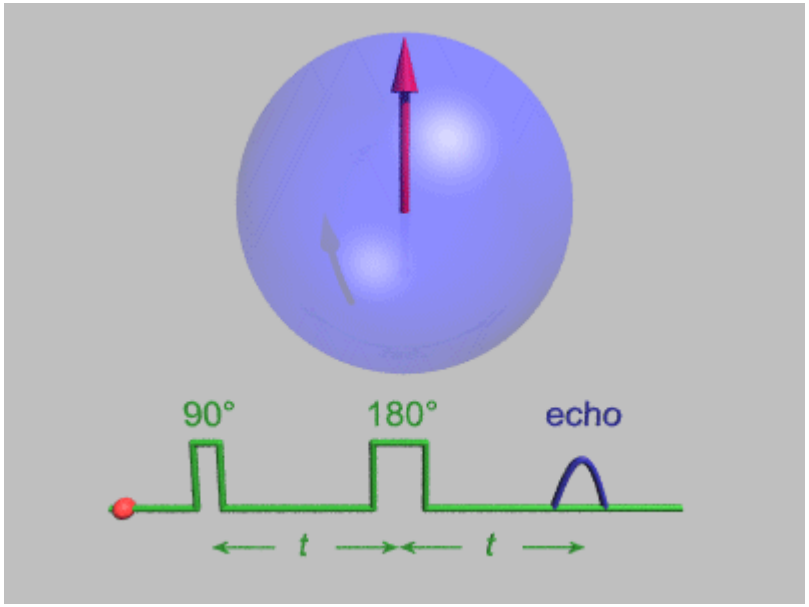


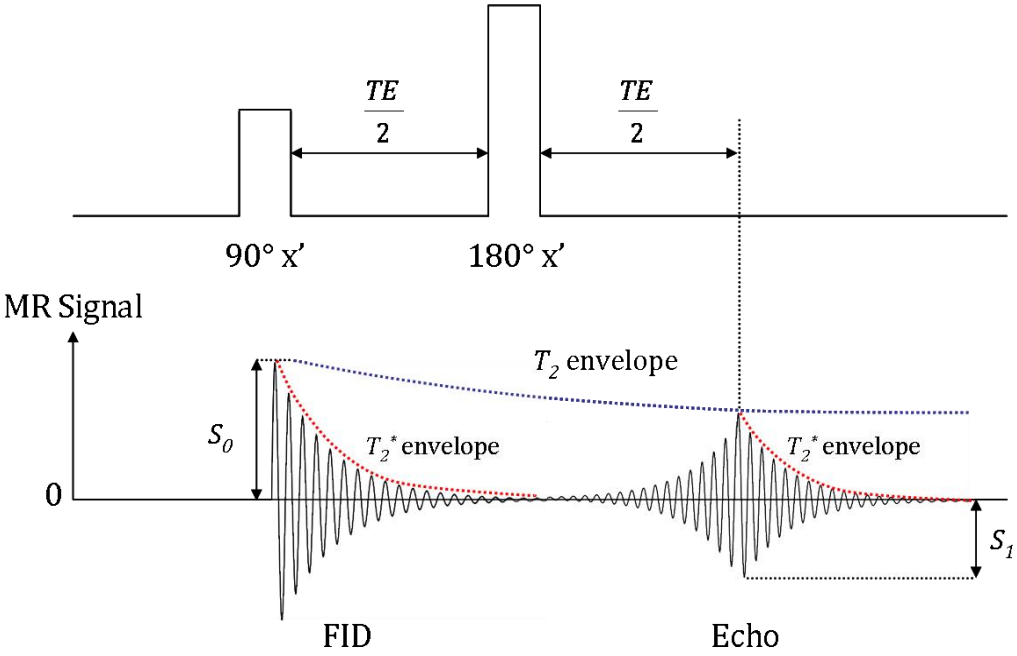


Transverse Decay (T_2 Relaxation)

* T_2 cannot be estimated by simply measuring the decay constant of the FID since the NMR signals decay faster than predicted effect of T_2 alone. This is because an extra dephasing originating from magnetic field inhomogeneity accelerates the FID.
* However, this factor can be removed with the application of a spin echo pulse sequence.

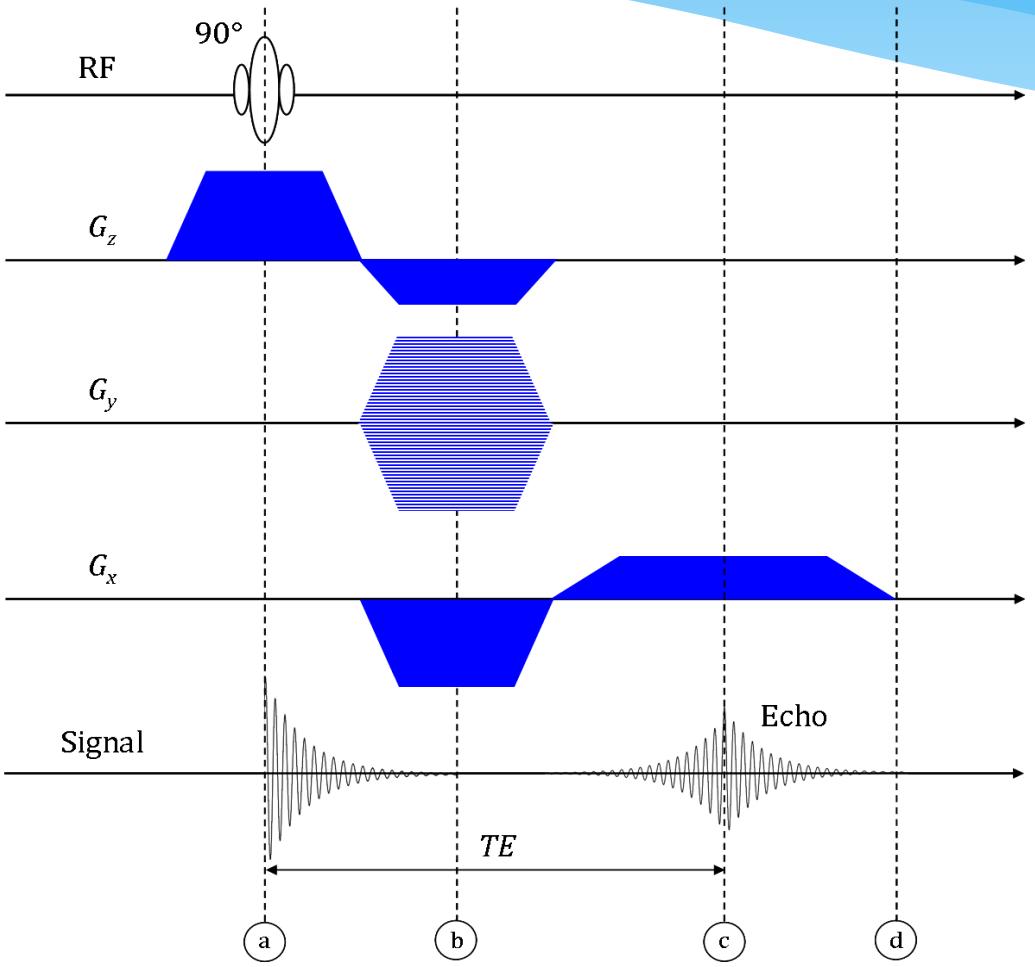
Spin Echo Pulse sequence

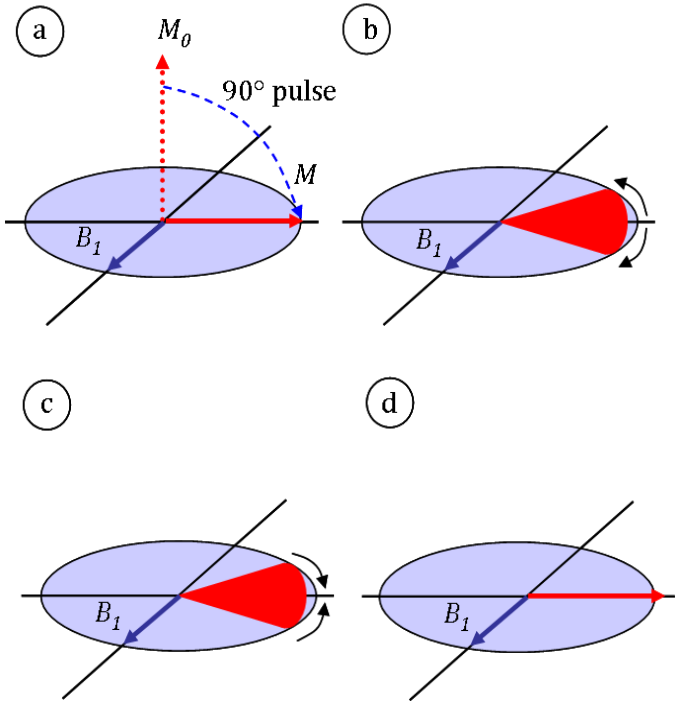




http://en.wikipedia.org/wiki/Spin_echo

Gradient Echo Pulse sequence





Spin Echo vs Gradient Echo

Choice of TR and TE for conventional spin echo sequences

TR	TE	
	Short (<40 ms)	Long (>75 ms)
Short (<750 ms)	T_1-weighted	Not useful
Long (>1500 ms)	PD-weighted	T_2-weighted

Choice of TR, TE and flip angle, α for gradient echo sequences

Flip angle, α	TE	
	Short (<15 ms)	Long (>30 ms)
Small (< 40°)	PD-weighted	T_2-weighted
Large (> 50°)	T_1-weighted	Not useful

Factors affecting Relaxation Times

* The T_1 and T_2 relaxation times can be affected by many intrinsic (tissue properties) and extrinsic (operator dependent) factors.
* Some main factors affecting the relaxation times are field strength, water content, viscosity, temperature and paramagnetic component of tissues.

T1 and T2 Relaxation Times

Tissue	T1 (0.5T)	T1 (1.5T)	T2
Fat	210 ms	320 ms	190 ms
Liver	350 ms	500 ms	40 ms
Muscle	550 ms	870 ms	45 ms
White Matter	500 ms	780 ms	90 ms
Grey Matter	650 ms	1100 ms	100 ms
Cerebrospinal Fluid (CSF)	1800 ms	2400 ms	190 ms

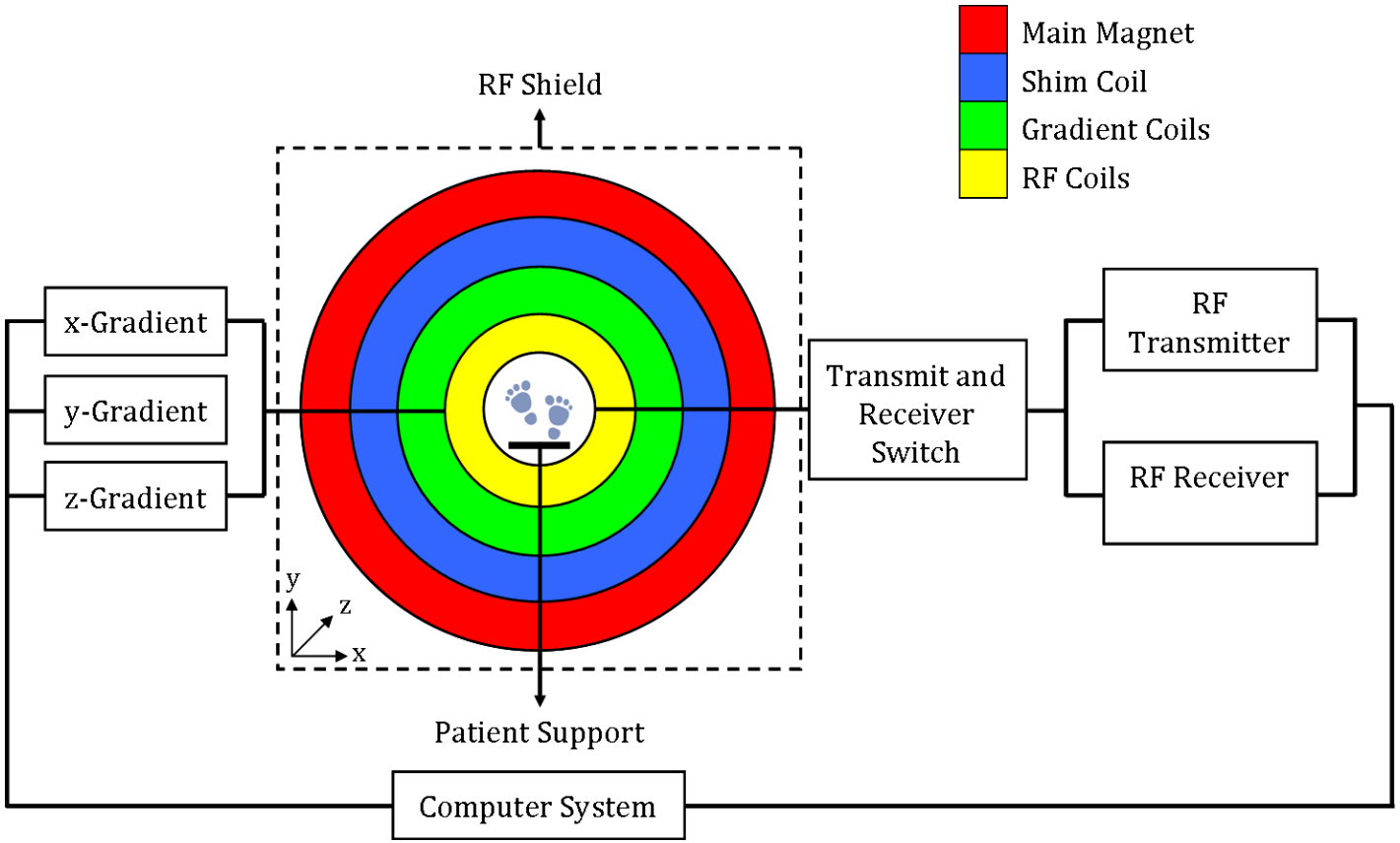
Functional Imaging

* fMRI attempts to localize the increased neuronal activation in the brain arising from specific tasks so as to produce maps of neuronal function.
* At fields of 3T or below, T_2^* weighted gradient echo, echo planar imaging is the most commonly used BOLD sequences.
* At 7T or higher, T2 weighted spin echo techniques are generally preferred.

Summary so far… section 2

* Actual MRI acquisition is quite complex. The T1 and T2 weighted images presented here involve more complex magnetisation manipulations and RF signal detection.
* However, we have covered the basic concept towards understanding how diagnostic MR images are generated.

Principle of MR Imaging



Magnetic Field Gradient

* Scanner has three orthogonal configured gradient coils → produce field gradients when current flows through loops of wires situated around the magnet's bore.
* Each gradient field generates its own magnetic field (G_x, G_y, G_z) that interacts with the main magnetic field (by either adding or subtracting).
* This leads to linear variation of the field and ω_L, as a function of distance from the centre of the magnet bore.

Magnetic Field Gradient

* The magnetic field gradient:
  * dephases and rephases the magnetic moments of nuclei
  * locates a slice within a selected plane of scan (slice selection)
  * spatially encodes the NMR signals of anatomy along the long (frequency encoding) and short (phase encoding) axis, as well as motion and flow encoding.
* Applications of these gradients are inter-changeable, so that imaging is possible at different angles including obliquely to the principle axes by linear combination of gradients.

Slice Selection

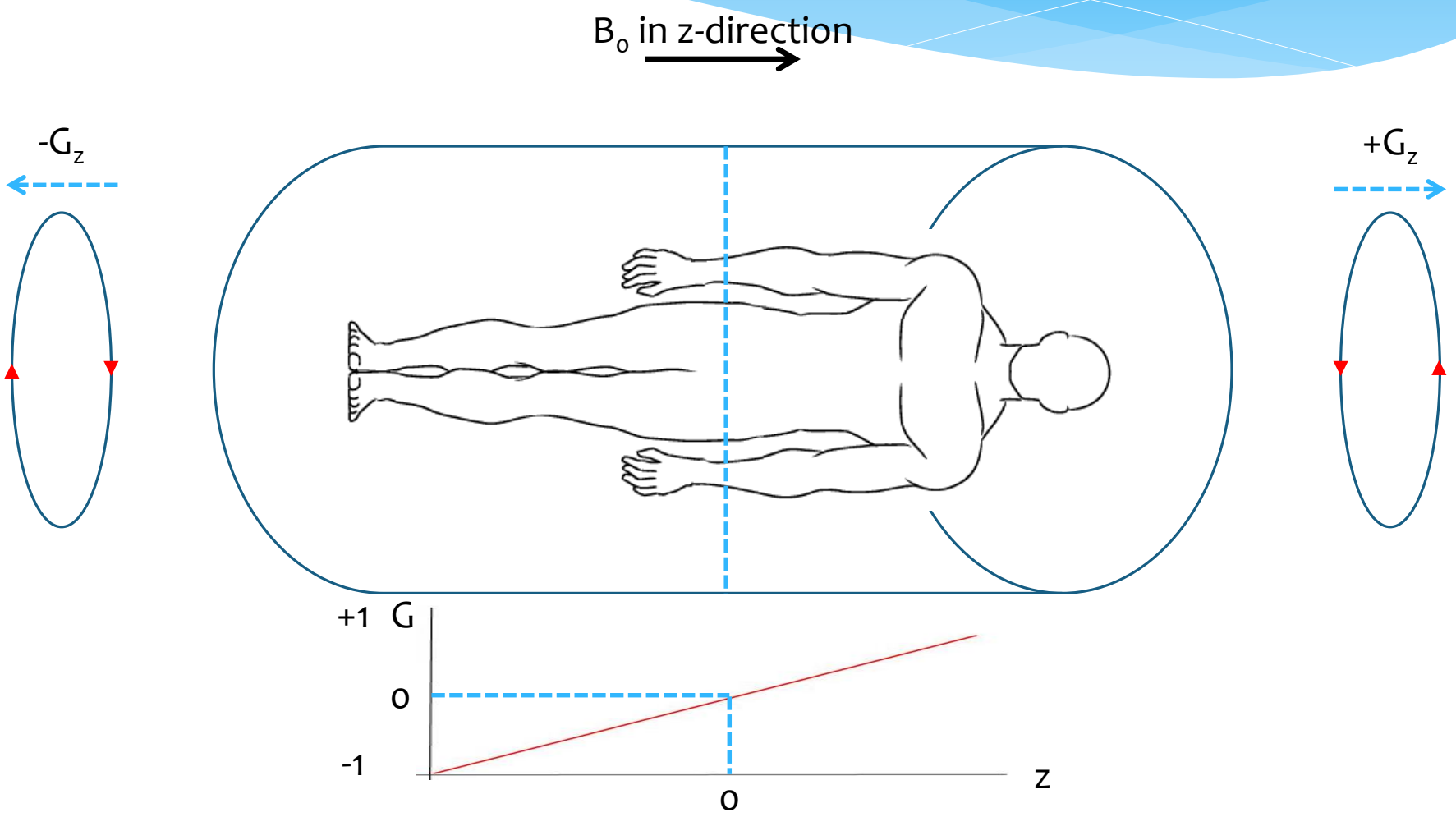
* Spatial selective radio frequency pulse can be employed for various function in MR imaging. It includes:
  * Excitation
  * Refocusing
  * Inversion
  * Spatial presaturation of magnetisation

  → which can be achieved through application of slice selective gradients

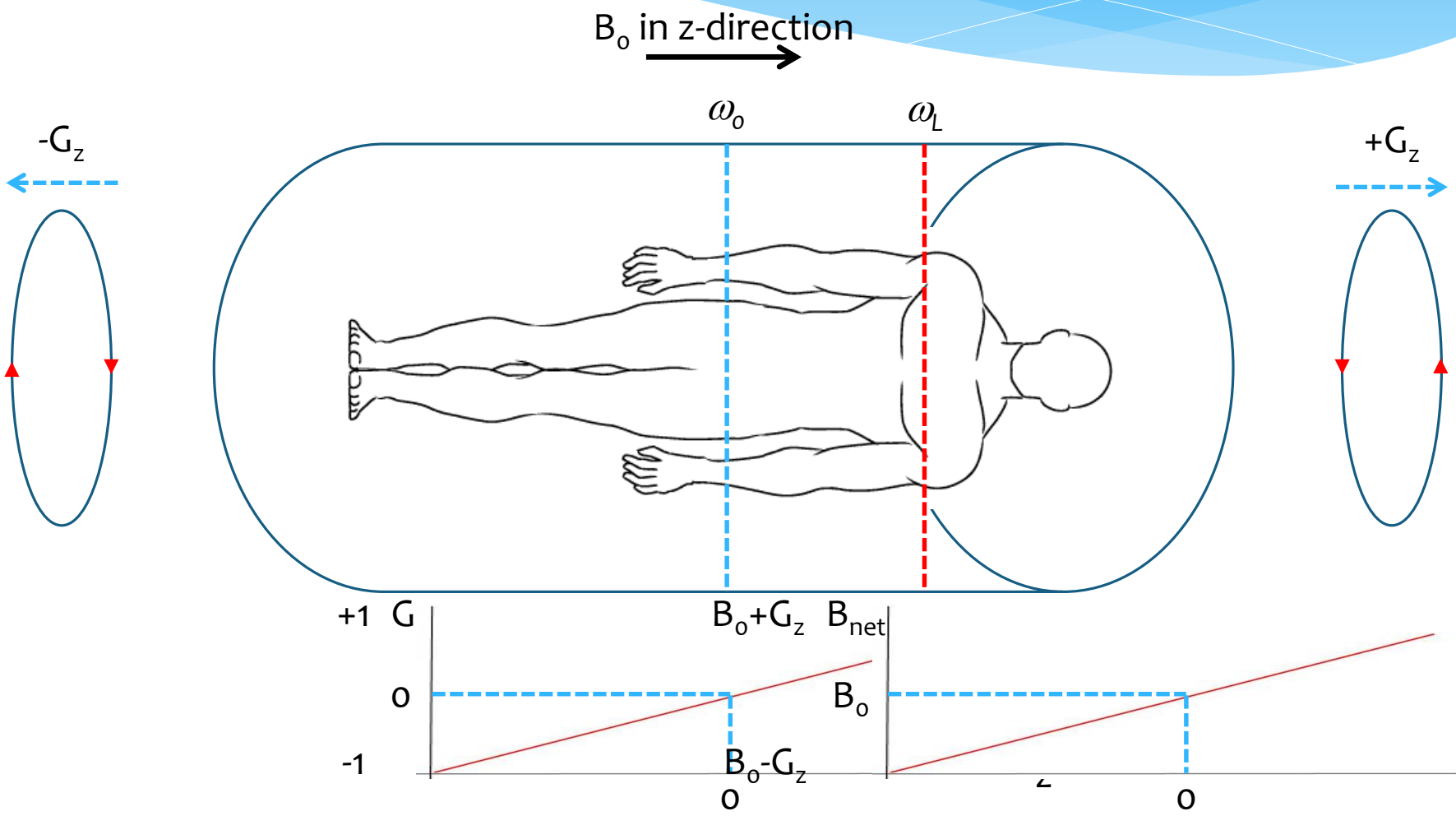
Slice Selection

* Slice selection is a technique to excite the spins in an isolated plane within the sample/patient that is being imaged.
* It is achievable with the application of an RF pulse that only affects a limited part of the sample in the presence of a linear field gradient along the direction of the desired slice.
* This procedure excites those spins (dominated by their position) with ω_L similar to the frequency of the applied RF pulse.

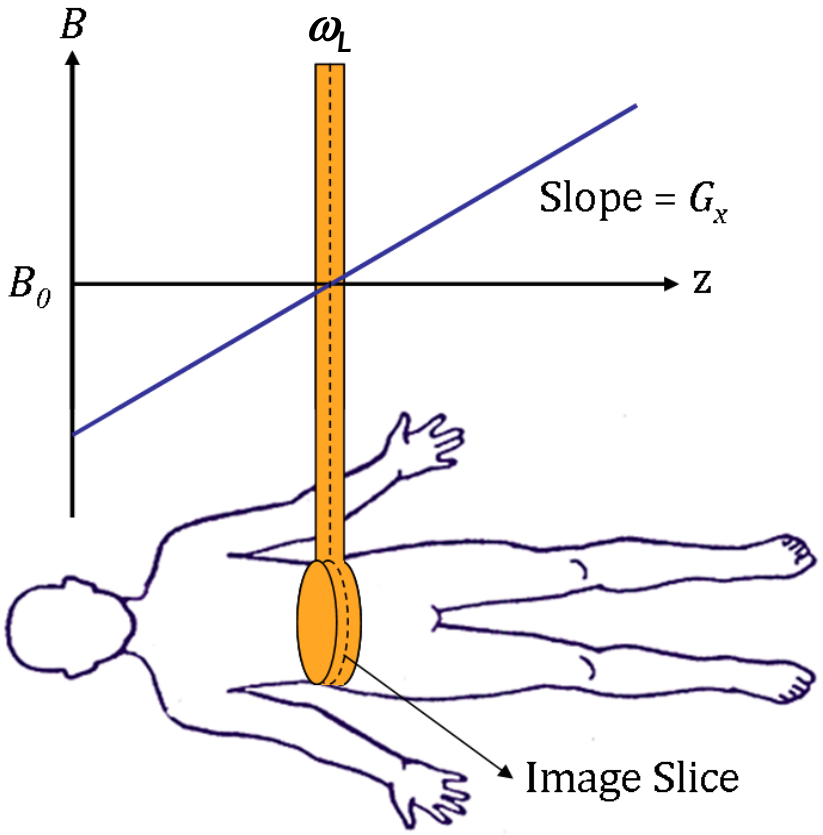
Slice Selection

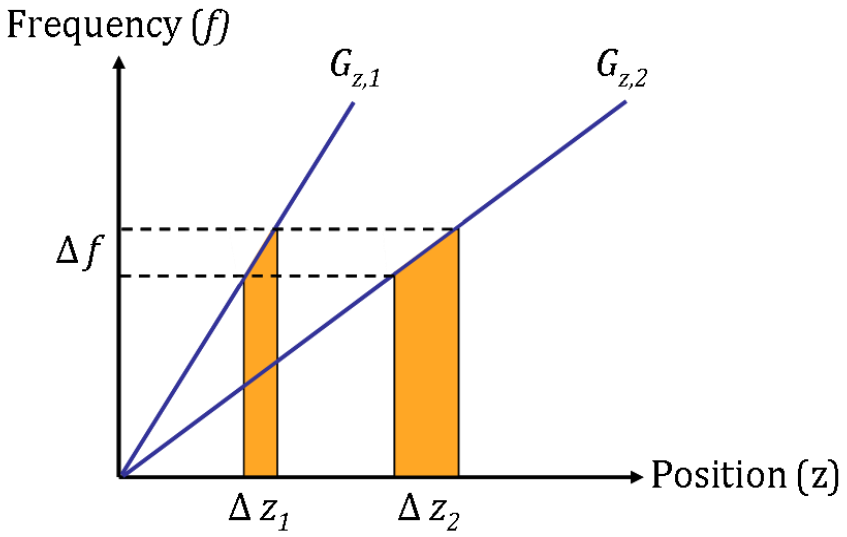


Slice Selection



Slice Selection





Spatial Encoding

* Following slice selection, the resulting signals do not have spatial information along the remaining two axes. Hence, the signals along these axes should be encoded using the spatial encoding technique that comprises two steps:
  * **frequency encoding**
  * **phase encoding**

Frequency Encoding

* Frequency encoding is a process of encoding signals along the long axis (x) of the anatomy.
* This can be achieved by switching on the frequency encoding field gradient while measuring the NMR signals (image readout).
* The application of a constant linear frequency encoding gradient along the x-axis will change the precessional frequency of the spins making up the signal.

$$\omega_L = \gamma (B_o + G_x x)$$

Frequency Encoding

* Since the resulting signal consists of a mixture of different frequency components, the signal needs to be sampled (digitised) using an ADC during the application of the frequency encoding gradient → done ASAP after removing the effect of the RF pulse. Then, the computer performs a Fourier transform.
* This results in a histogram demonstrating the quantity of signal present at a number of discrete, equally spaced frequency ranges.

Frequency Encoding

* By using the Fourier transform, the position can be encoded into frequency resulting in a 1D projection of the region of interest in the direction of the frequency gradient.

Phase Encoding

* Phase encoding localises the NMR signals orthogonal to the frequency encoding direction, to generate a 2D image. The G_y is usually applied just before sampling the signal.
* Following slice selection and frequency encoding, NMR signals have been located to a specific slice within a desired plane encoded along a pre-determined direction. The spins in the pixels of the same column are all in phase since all of these spins precess at the same frequency.

Phase Encoding

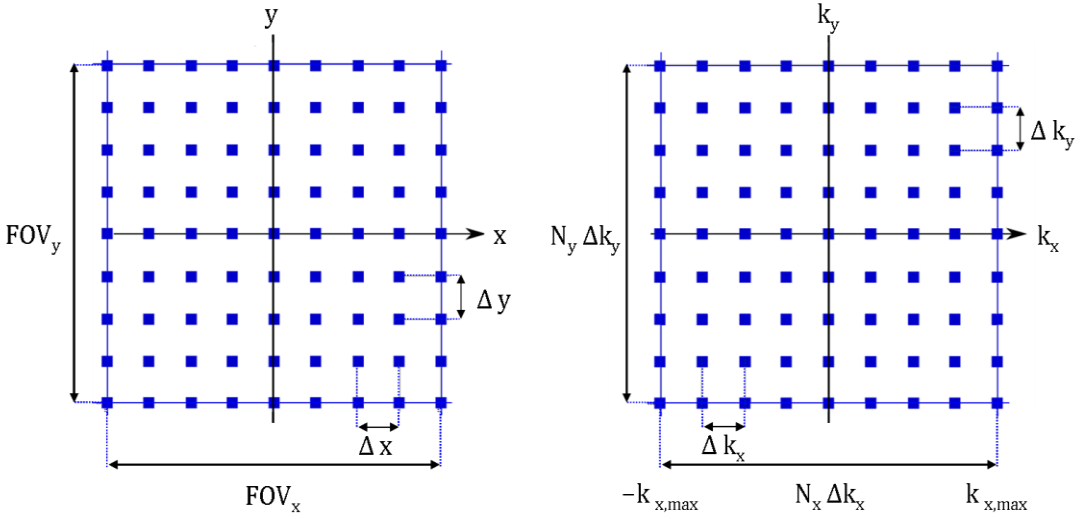
* Phase cannot be measured directly. However, phase difference can be measured!
* In contrast to frequency encoding that samples all of the data following a single RF excitation pulse, phase encoding needs to be repeated a number of times corresponding to the number of k-space lines that needs to be sampled.
* The resulting signals can be reconstructed using Fourier transformation to restore the spatial information of the sample or patient.

Fourier Transform

* Complex signals can be rewritten as an infinite sum of simple sinusoidal waves (in terms of simple sines and cosines).
* The Fourier Transform decomposes a complicated signal into amplitude and frequencies of its simple wave components.
* In MRI, this concept is used to encode the frequency and phase encoding information of the signals within k-space to construct MR images.

k-Space

* k-space is used to visualise the trajectories of the spins' phase under the influence of field gradients. The k-space of a two-dimensional: horizontal axis (frequency encoding axis) and vertical axis (phase encoding axis)



Summary so far… section 3

* Magnetic field gradients form the basis of MR signal localisation.
* 2D slices are produced by a combination of an excitation RF pulse and simultaneous slice-selection gradient.
* Due to z-gradient → slice selection
* Due to x-gradient → different frequencies
* Due to y-gradient → different phases
* We sample every spatial frequency that can exist within the image before we Fourier Transform these data (known as k-space) to produce the image directly.

Recommended Reading

* D.W. McRobbie, E.A. Moore, M.J. Graves, and M.R. Prince. (2003), MRI From Picture to Proton, Cambridge University Press, Cambridge, United Kingdom.
* M.A. Bernstein, K.F. King, and X.J. Zhou. (2004), Handbook of MRI Pulse Sequences, Elsevier Academic Press, USA.
* C. Westbrook, C.K. Roth, and J. Talbot. (2005), MRI In Practice, Blackwell Publishing Ltd, Oxford
* E.J. Rummeny, P. Reimer, and W. Heindel. (2006), MR Imaging of the Body, Thieme, Stuttgart, Germany.

Recommended Video

* Introducing MRI lecture series from Albert Einstein College of Medicine: https://www.youtube.com/watch?v=35gfOtjRcic&list=PLPcImQzEnTpz-5TzxyyoYSbiAa9xdd89l
* Introductory NMR and MRI with Paul Callaghan: https://www.youtube.com/watch?v=7aRKAXD4dAg&list=PL2955748835024861
* How does an MRI work? https://www.youtube.com/watch?v=pGcZvSG805Y