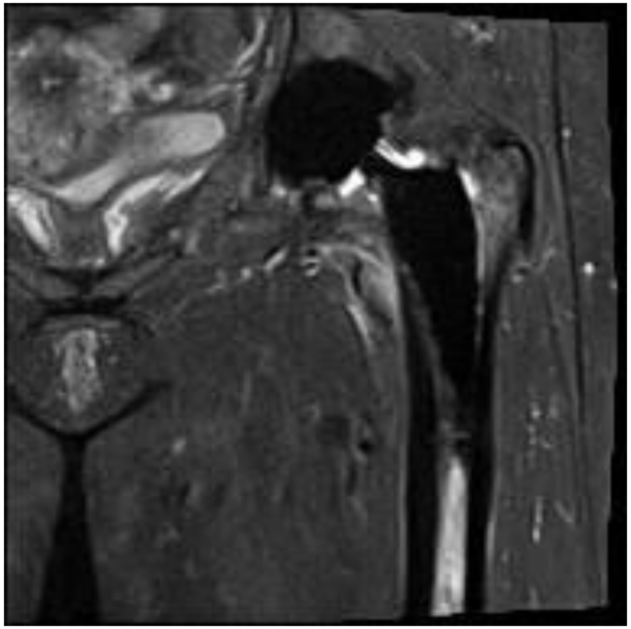
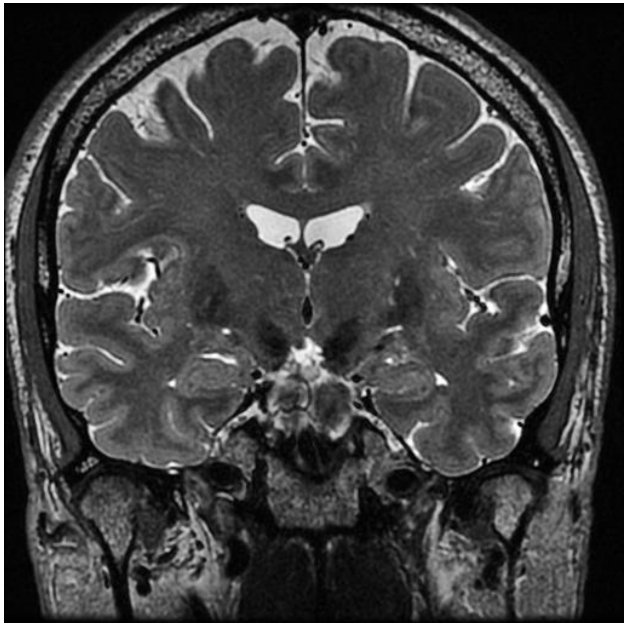
Magnetic Resonance Imaging

F.R.C.R. Physics Lectures

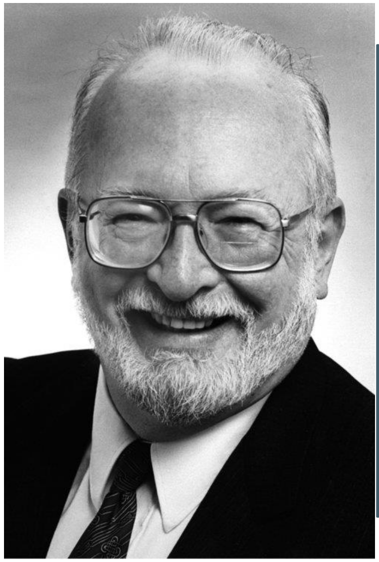




Lawrence Kenning PhD

History

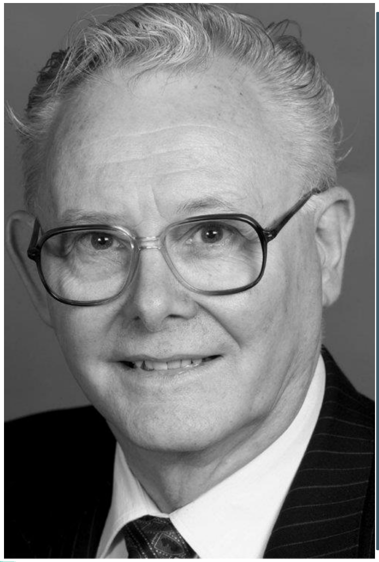
- 2003 Nobel Prize in Physiology or Medicine was awarded to Paul Lauterbur and Peter Mansfield for their discoveries concerning magnetic resonance imaging (MRI)
- Lauterbur's imaging experiments moved science from the single dimension of NMR spectroscopy to the second dimension of spatial orientation - the foundation of MRI
- Mansfield further developed the utilisation of gradients in the magnetic field. He showed how the signals could be mathematically analysed, which made it possible to develop a useful imaging technique



Paul C. Lauterbur

The Nobel Prize in Physiology or Medicine 2003

1929 - 2007



Sir Peter Mansfield

The Nobel Prize in Physiology or Medicine 2003

1933 - 2017

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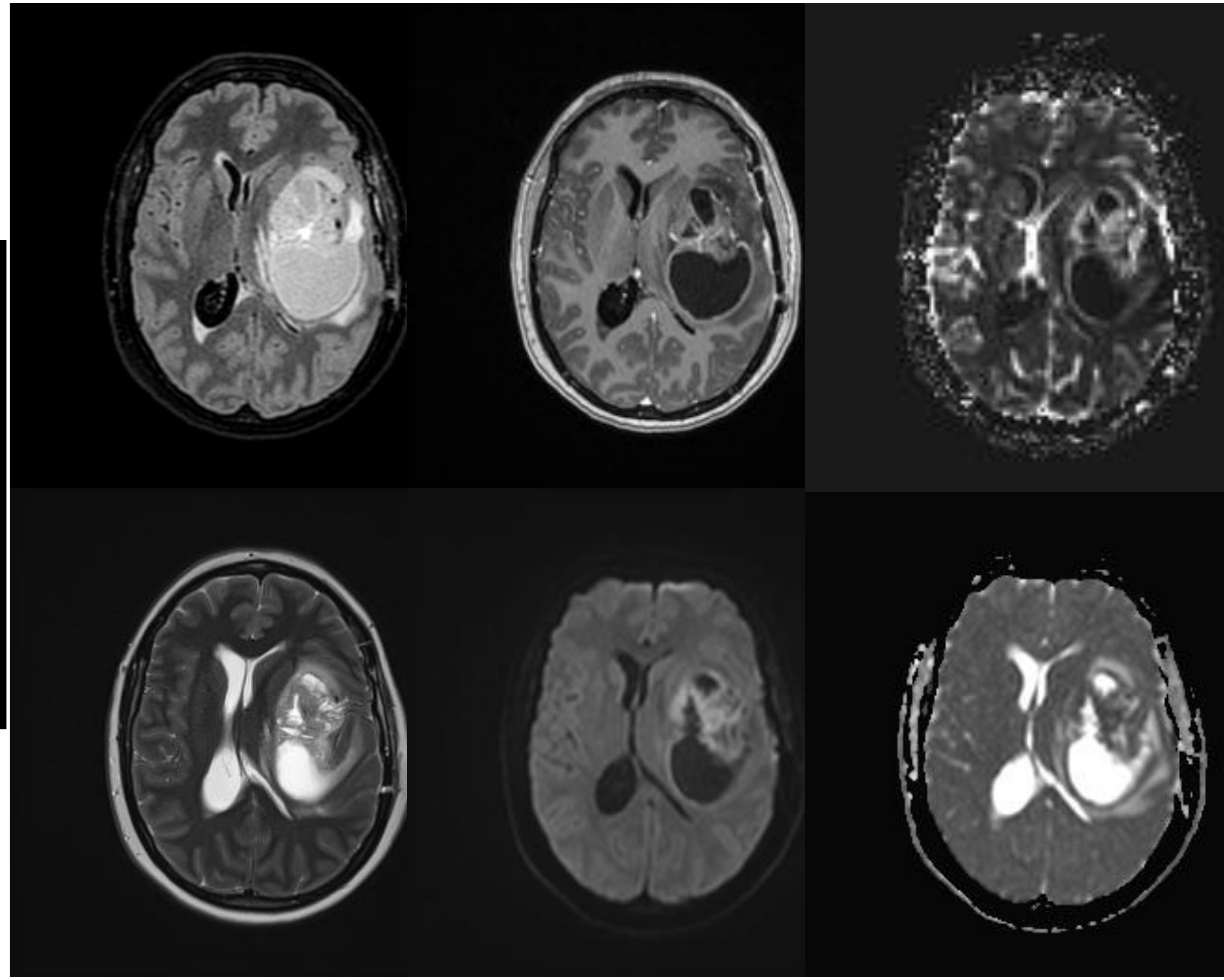
MRI:

- Non-ionising radiation. Suited to patients requiring multiple imaging examinations
- Large range of tissue contrasts. Superior soft tissue contrast.
- Functional imaging ‘flavours’ such as diffusion, perfusion and metabolism
- Scanning can be performed in any imaging plane

CT:

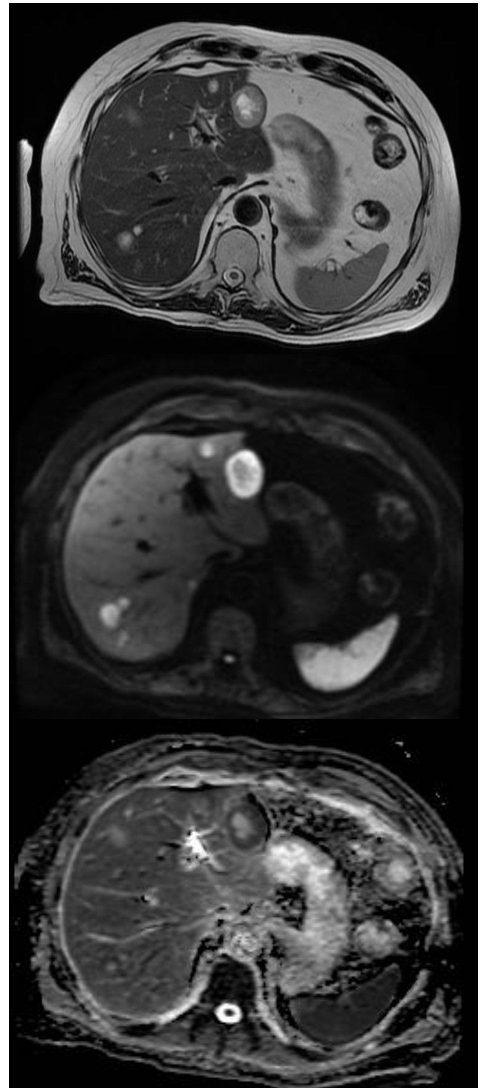
- Much faster than MRI (less sensitive to motion)
- Cheaper than MRI
- Accurate detection of calcification and metal foreign bodies
- No risk to the patient with implantable medical devices

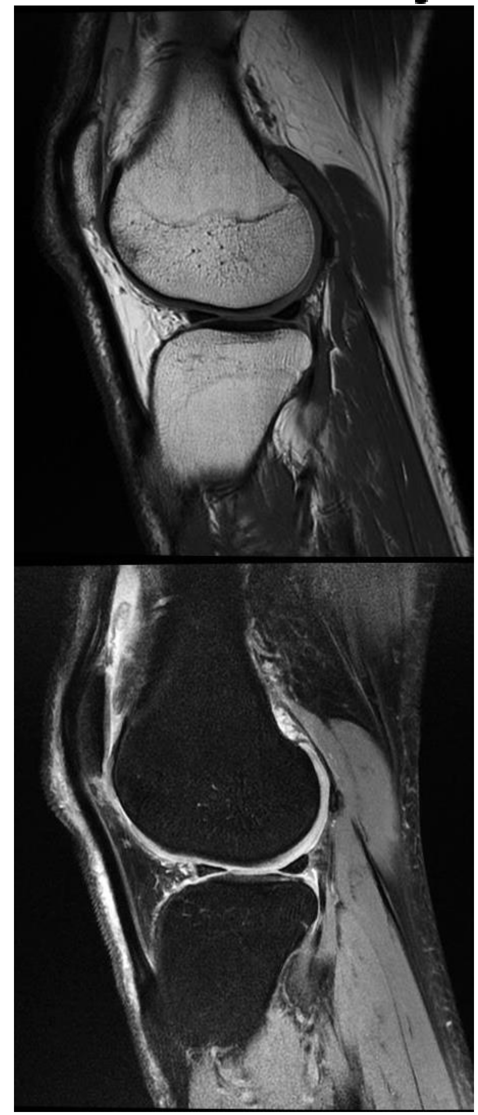
Soft Tissue Contrast



Today

- Neurology
- Musculoskeletal
- Cardiac
- Gastrointestinal
- Oncology

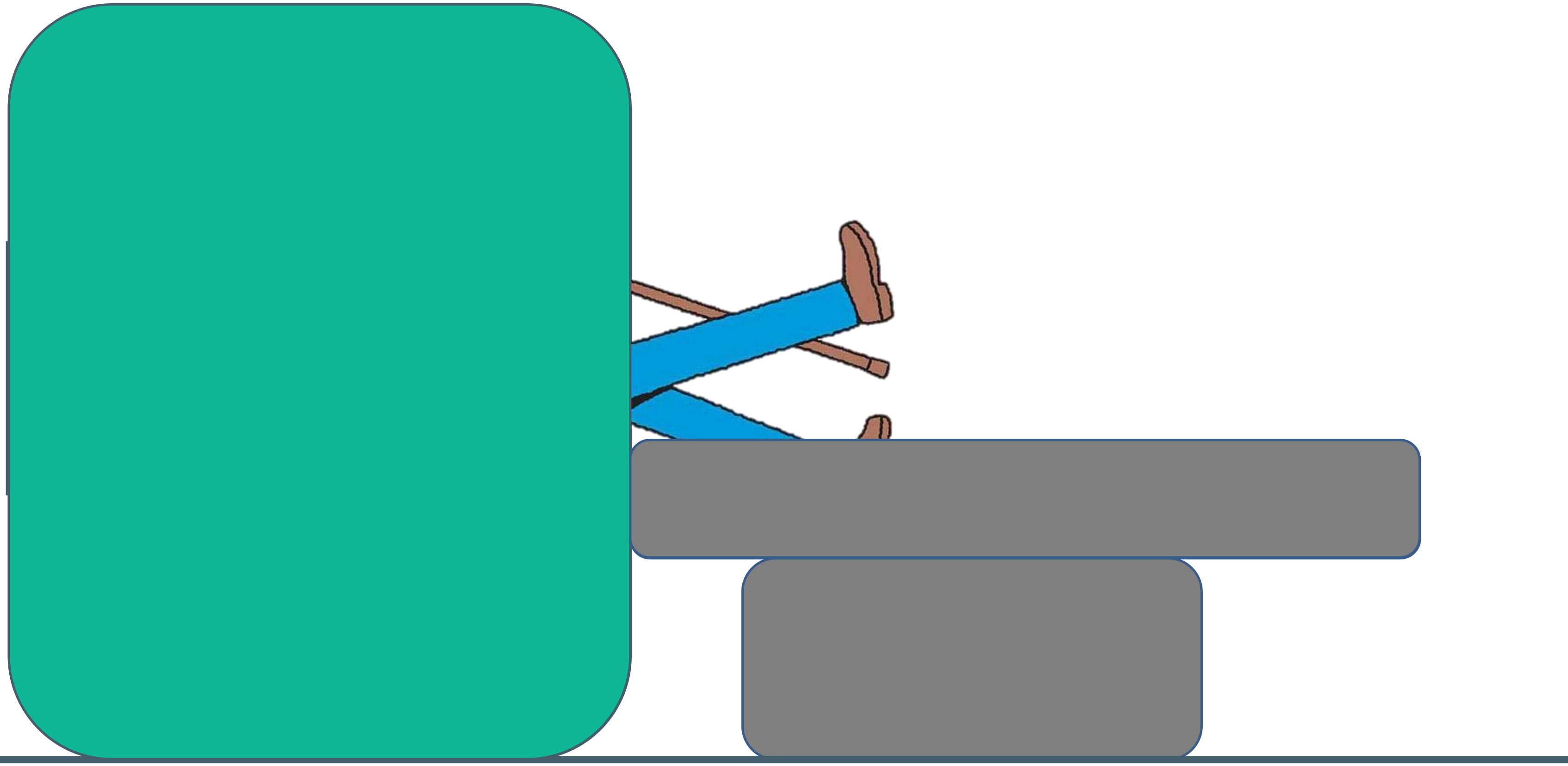




MRI – A Brief Summary

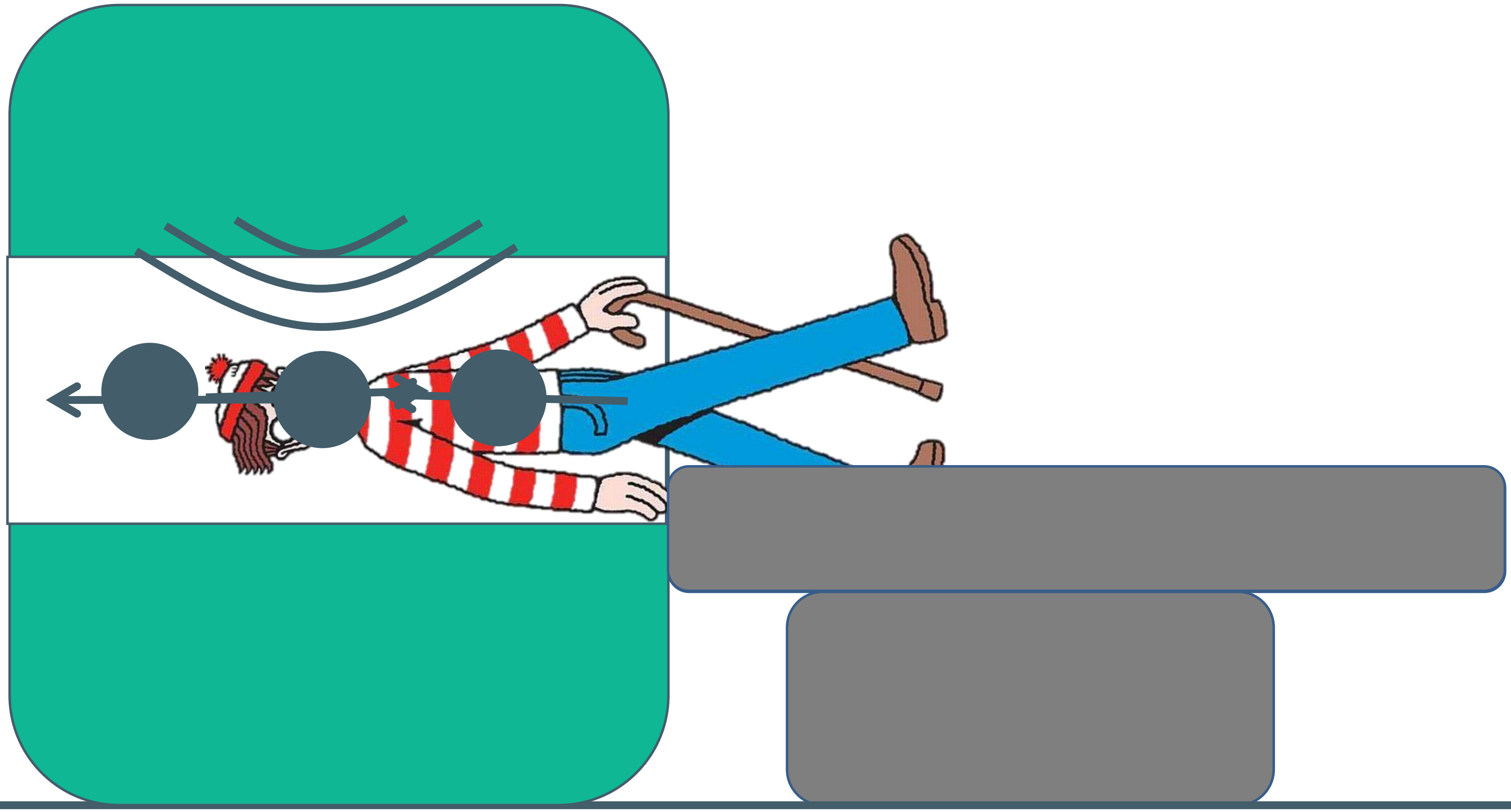
- Magnetic resonance imaging consists of placing the patient inside a large magnetic field

- Magnetic field causes protons in water molecules to align with/against the magnetic field



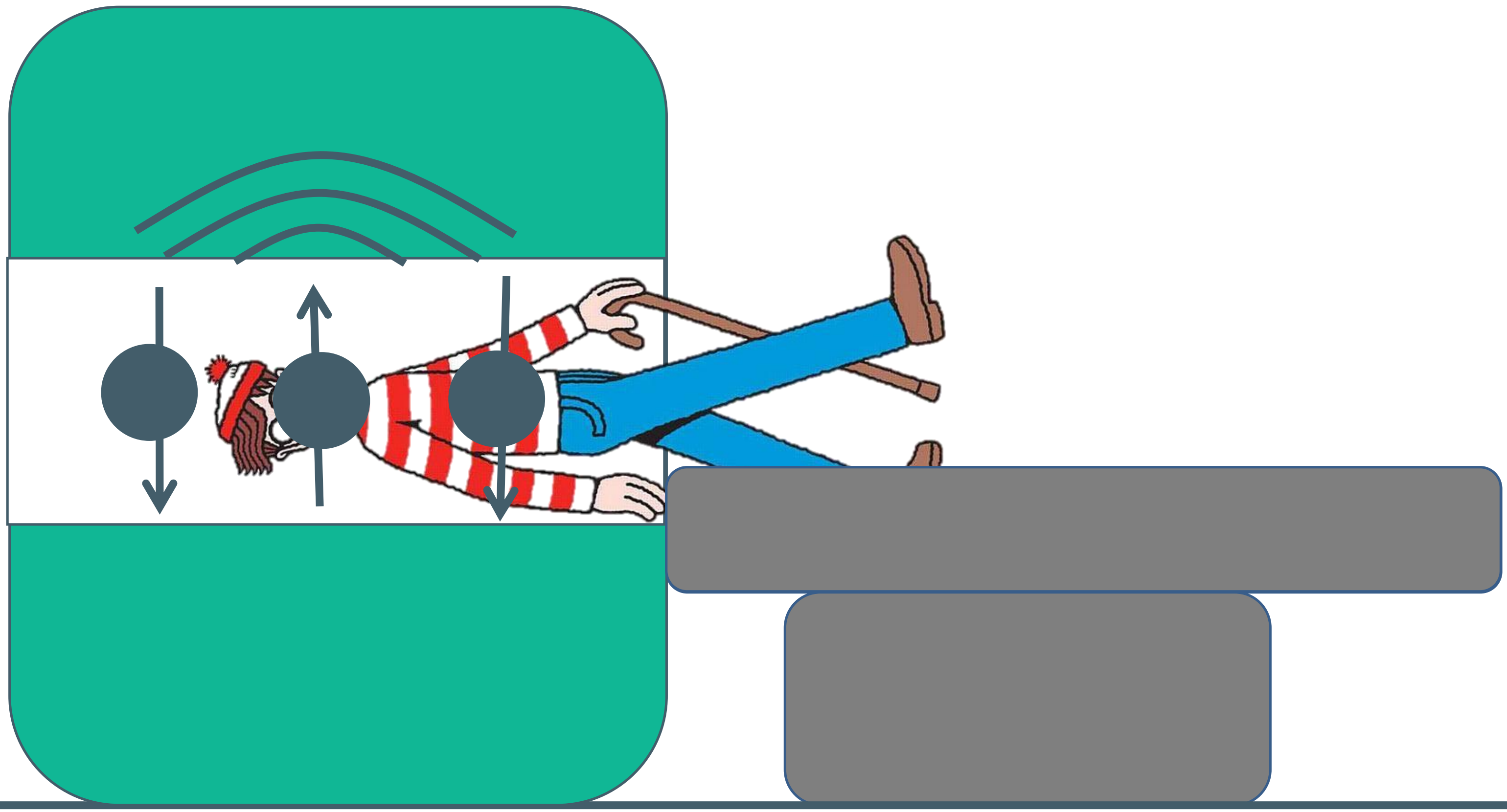
MRI – A Brief Summary

- Radiofrequency (RF) pulses are used to “excite” protons



MRI – A Brief Summary

- Energy is subsequently released as RF signal



MRI – A Brief Summary

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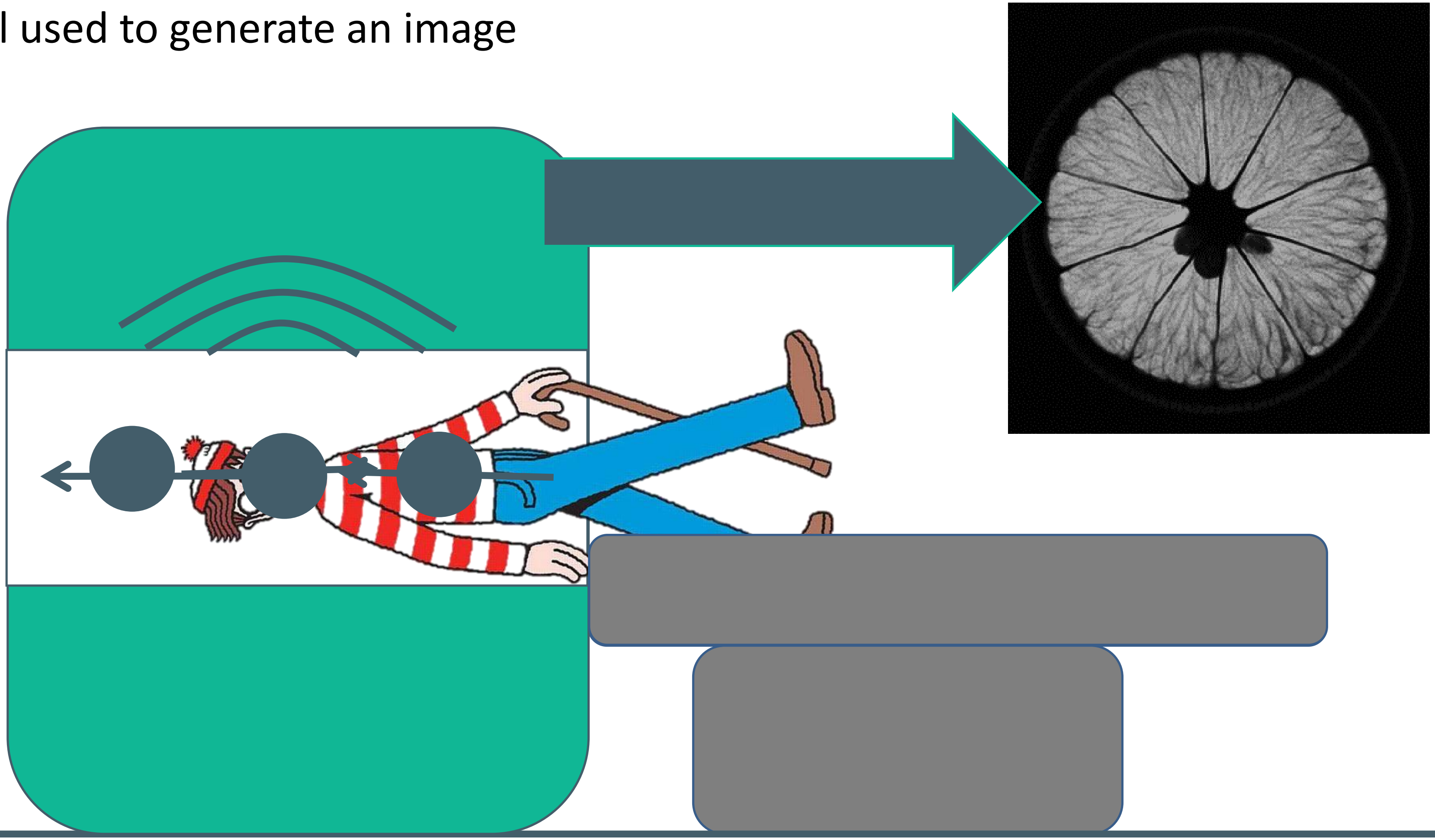
- Signal used to generate an image



Creation, detection and spatial localisation of MR signal

7.1 Creation, detection and spatial localisation of MR signal

- Nuclear magnetic resonance
- Precession about magnetic fields (B_0 and B_1)
- Equilibrium magnetisation (M_0) and dependence on the strength of the magnetic field, B_0
- Longitudinal (M_z) and transverse magnetisation (M_{xy})
- Slice Selection
- k-space:
  - Relationship between k-space and MR image
  - Frequency-encoding
  - Phase-Encoding
  - Awareness of different k-space trajectories and their advantages/disadvantages
- 2D versus 3D sequences

NMR to MRI

- MRI is based on the phenomenon of Nuclear Magnetic Resonance (NMR)

- NMR does exactly what it says on the tin
  - Related to the nucleus
  - Related to magnetism and magnetic fields
  - Related to resonant behaviour

- When imaging was added, it became NMRI and then MRI
- "Nuclear" was dropped to avoid negative associations

Nuclear Spin

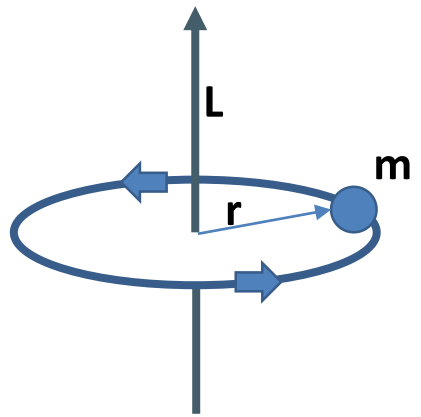
- In classical physics, a rotating "spinning" object possesses a property known as **angular momentum**. Angular momentum is a form of inertia, reflecting the object's size, shape, mass, and rotational velocity. It is typically represented as a vector (**L**) pointing along the axis of rotation.

- Consider a spinning ring. Spin will give this object an "angular momentum" **L**, which are related though mass (m), radius (r) and angularly velocity (ω) in the following equation:

$$\mathbf{L = m\, r^2\, \omega}$$

$\mathbf{I = m\, r^2}$ **(moment of inertia for a ring)**

$$\mathbf{L = I\, \omega}$$



Nuclear Spin

- Atomic particles posses a corresponding property known as **spin** or **spin angular momentum.** Protons, neutrons and electrons all possess **spin**, however, several key differences should be noted:
  - The particle is **not actually spinning** or rotating
  - Spin, like mass, is a fundamental property (purely **quantum mechanical quantity**)
  - **Spin interacts with electromagnetic fields** whereas classical angular momentum (L) interacts with gravitational fields
  - The magnitude of spin is **quantised**, meaning that it can only take on a limited set of discrete values

- Nuclear spin values range from I = 0 to I = 8 in ½-unit increments

- The physical basis of Nuclear Magnetic Resonance (NMR) centres around the concept of a nuclear **spin**, its associated angular momentum (**s**) and its magnetic moment (**μ**)

Nuclear Spin

- Consider a proton or hydrogen (^{1}H) nucleus. Spin will give this nucleus a "spin angular momentum," **s**, and a magnetic moment, **μ**, which are related though a proportionality constant, **γ**, in the following equation:

$$\boldsymbol{\mu} = \boldsymbol{\gamma}\, \mathbf{s}$$

- **s** and **μ** are vector quantities and like many things in quantum mechanics, they can only take on discrete values.

- The spin (**I**) of different nuclei are interpreted using the nuclear shell model

- The component m_I of nuclear spin parallel to the z–axis can have (2I + 1) values

Nuclear Spin

- ^{14}N nucleus has I = 1, so that there are 3 possible orientations relative to the z–axis, corresponding to states m_I = +1, 0 and −1
- Even-even nuclei with even numbers of both protons and neutrons, such as ^{12}C and ^{16}O, have spin zero
- Odd mass number nuclei have half-integer spins, such as 1/2 for ^{1}H, 3/2 for ^{7}Li and 1/2 for ^{13}C
- Odd-odd nuclei with odd numbers of both protons and neutrons have integer spins, such as 3 for ^{10}B, and 1 for ^{14}N.

Nuclear Spin

- If non-zero, nucleus has a magnetic moment - can detect with magnetic resonance.

- Spin = 0 nuclei *cannot* generate NMR signals.

- ^{1}H ✓
- ^{12}C ✗
- ^{14}N ✗
- ^{16}O ✗

Nuclear Spin

- Many nuclei are NMR observable and the following are "clinically" relevant
- However, ^{1}H is the dominant nucleus in MRI

Nucleus	Spin	γ Rad/T/s ($\times 10^{8}$)	Frequency @1.5T (MHz)
^{1}H	1/2	2.675	63.8
^{13}C	1/2	0.673	16.0
^{19}F	1/2	2.517	59.8
^{23}Na	3/2	0.708	16.8
^{31}P	1/2	1.083	25.7

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Philips MR 7700 3T

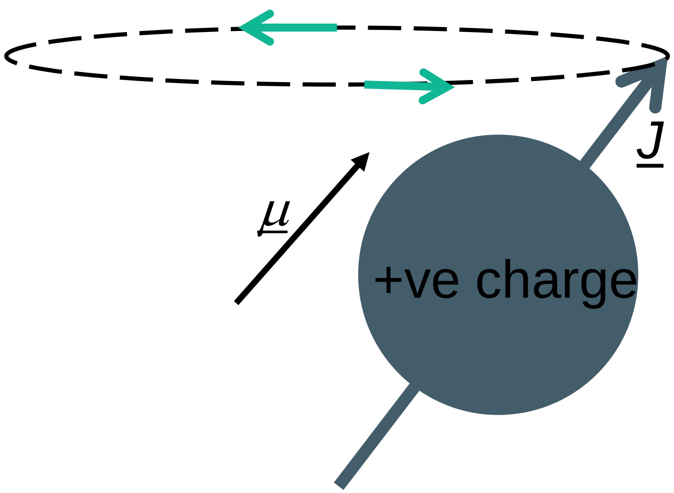
Imaging and spectroscopy of six different nuclei are possible using Philips MR 7700 3T:

- 1H
- 31P
- 13C
- 23Na
- 19F*
- 129Xe*

*Caution: Investigational device for imaging with fluorine (19F) and xenon (129Xe). Limited by federal (or United States) law to investigational use. Clinical imaging with these nuclei requires usage of a cleared drug. No FDA-cleared drugs are currently available for these nuclei.

Nuclear Spin

- The human body predominantly consists of water and fat, with hydrogen (m_I=1/2) as the main constituent (62% percent of atoms)
- Each ^{1}H atom consists of a single proton with an orbiting electron, with the nucleus then rotating around its own axis
- The rotating motion of the positively charged proton creates a magnetic field also known as a magnetic moment



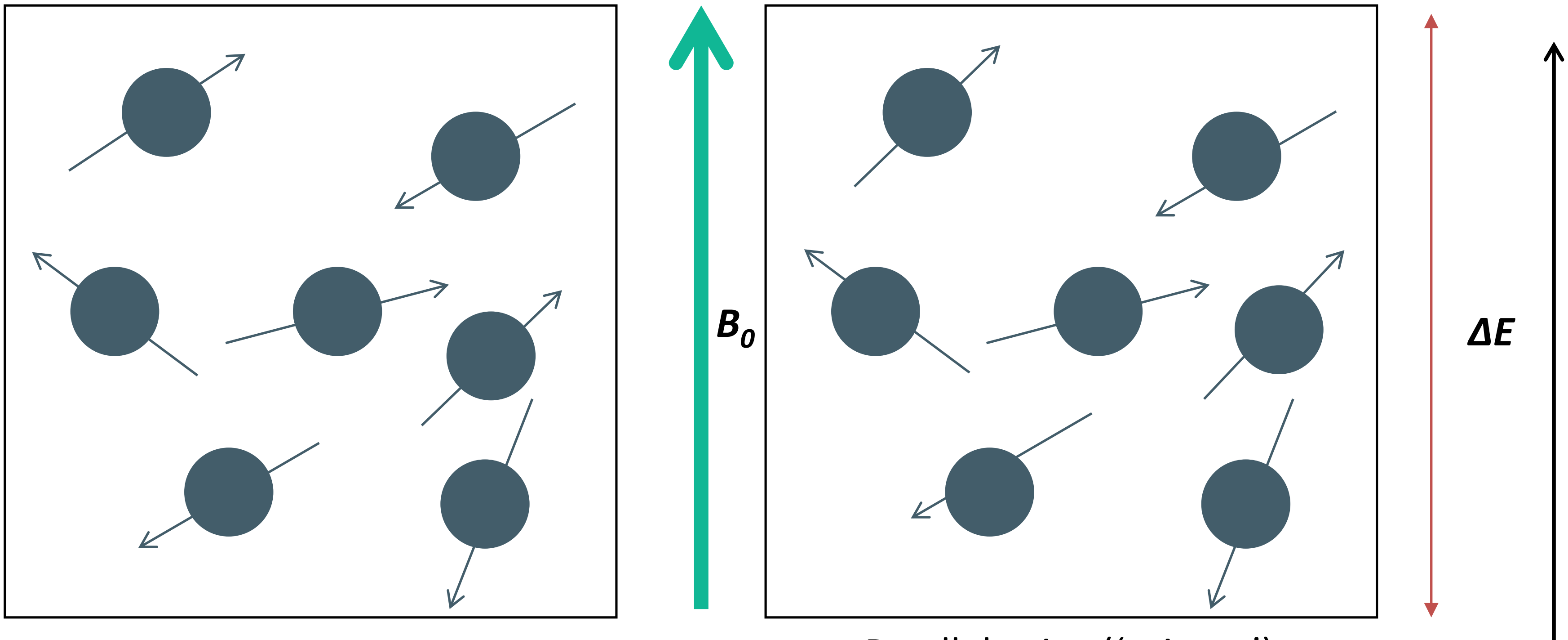
$$\underline{\mu} = \gamma \underline{J}$$

μ = magnetic moment. J = angular momentum. γ is the gyromagnetic ratio (constant for a given nucleus)

Equilibrium magnetisation (M_0) and dependence on the strength of the magnetic field, B_0

Charge in a field – bulk magnetisation

Antiparallel spins (‘spin down’)
Higher energy



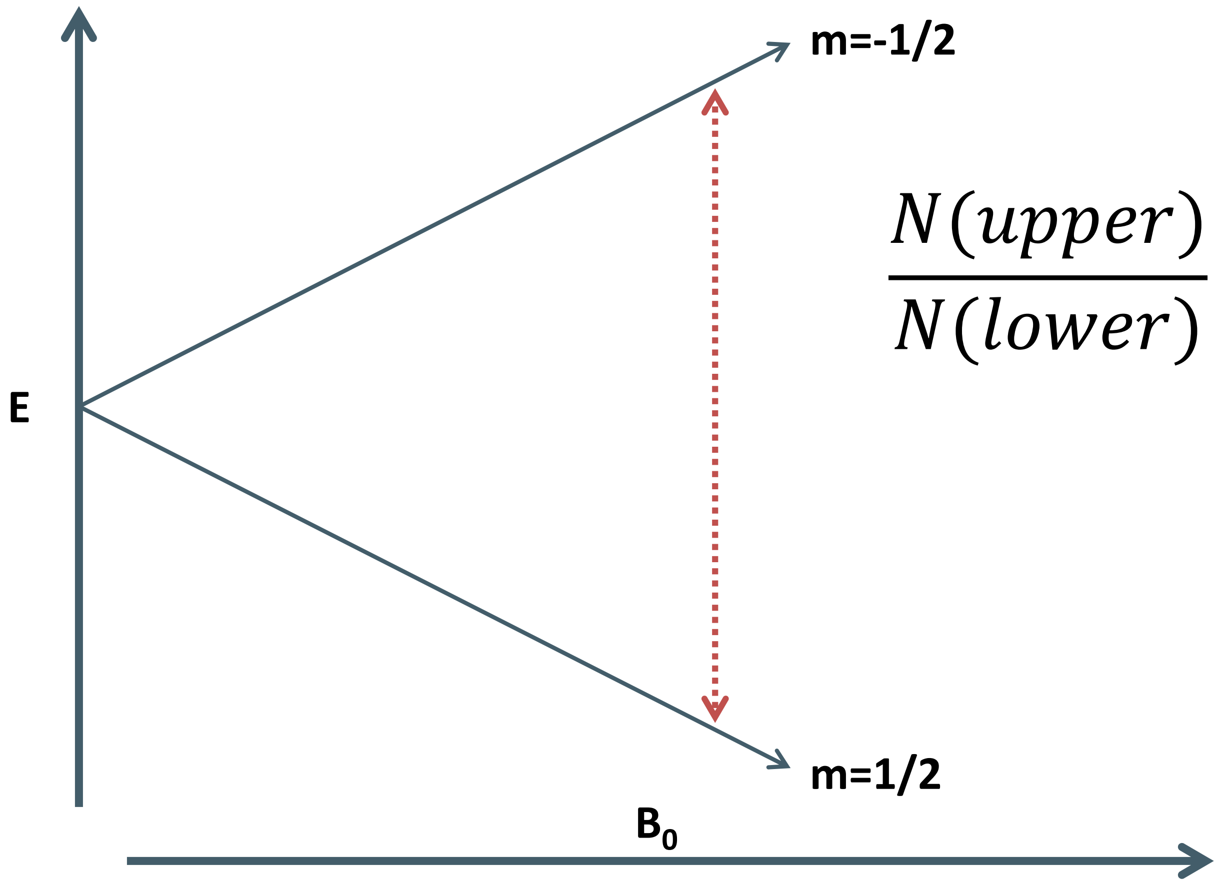
Parallel spins (‘spin up’)
Lower energy

***Net magnetic Moment* M_0**

Due to thermal energy at equilibrium, a slight majority exists in the low-energy, parallel direction. **B_0 = static magnetic field**

Equilibrium magnetisation (M_0) and dependence on the strength of the magnetic field, B_0

Boltzmann distribution



$$\frac{N(upper)}{N(lower)} = \exp(-\frac{\Delta E}{kT})$$

$$\Delta E = \frac{\gamma h B_0}{2\pi}$$

An increase in spin population difference can be achieved by increasing the magnetic field strength (B_0) or decreasing temperature (T)

m is the magnetic quantum number. *h* is Planck's constant
k is the Boltzmann constant (1.381 x 10^{-23} joules/°K). *γ* is the gyromagnetic ratio

Effect of field strength

- Stronger magnetic fields increase:
  - energy separation of the low- and high-energy levels
  - number of excess protons in the low-energy state

- At 1.0 T, the number of excess protons in the low-energy state is approximately 3 protons per million ($3x10^{-6}$)

- Typical voxel in MRI has 10^{21} protons. This means there are approximately $3x10^{15}$ more protons in the low-energy state!

- This number of excess protons produces an observable "sample" nuclear magnetic moment

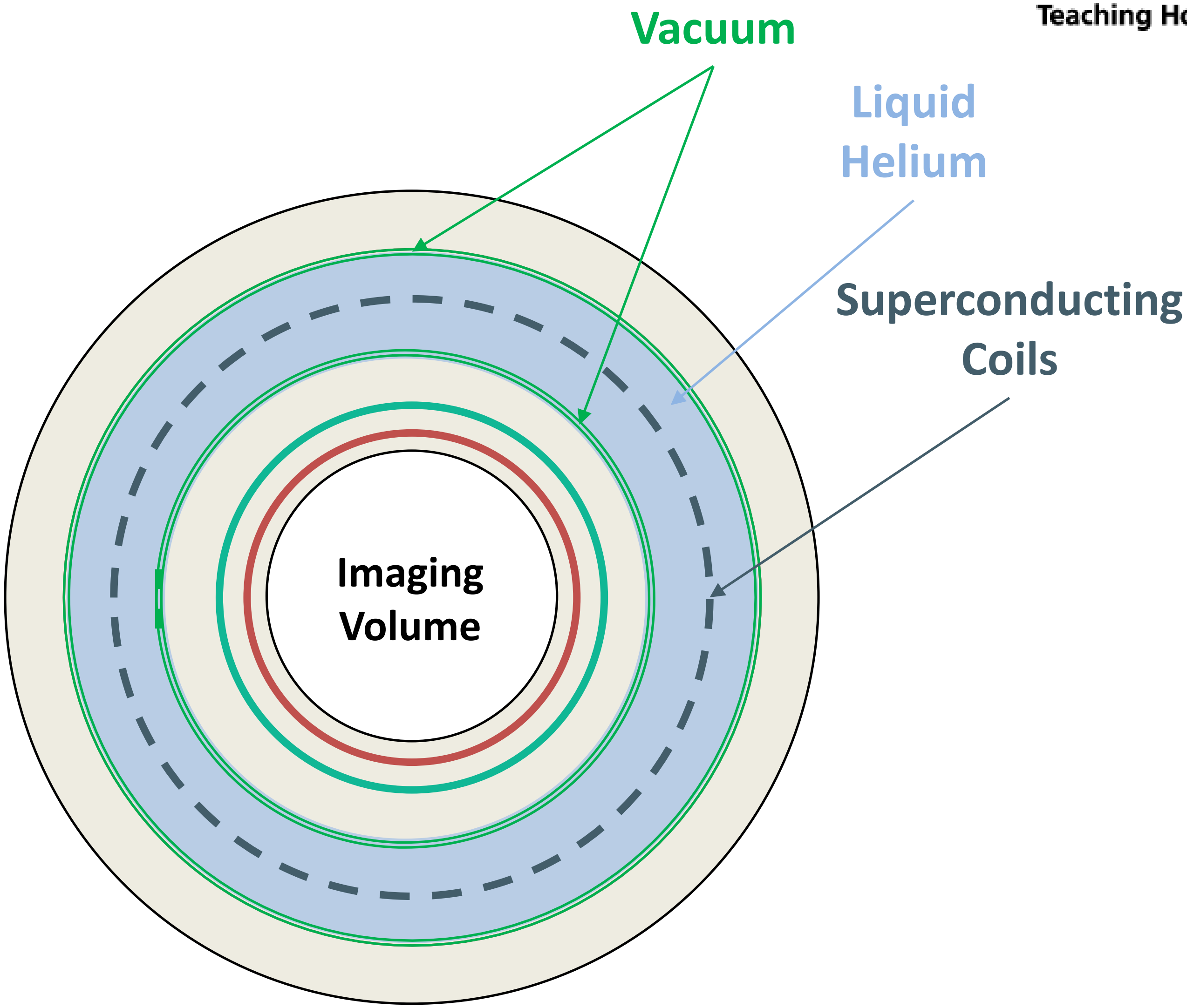
Static magnetic field (B_0)

- **High magnetic field strengths (>1 T)** requires electromagnet core wires to be superconductive
- Superconductivity is a characteristic of certain metals (e.g., niobium–titanium alloys) that when maintained at extremely low temperatures (liquid helium; **<4°K**) exhibit no resistance to electric current
- Superconductivity allows the closed-circuit electromagnet to be energised and ramped up to the desired current and magnetic field strength by an external electric source. This results in low power consumption once at field.
- Superconductive magnets with field strengths of 1.5 to 3.0 T are common for clinical systems, 7.0 T is used predominantly research but starting to become clinical
- High field homogeneity, High SNR, High capital costs (~£1M)

Static magnetic field (B_0)

Equilibrium magnetisation (M_0) and dependence on the strength of the magnetic field, B_0

System Design



Magnetic field strength

- The Earth's magnetic field ~ 0.5 Gauss
- 1 Gauss = 1×10^{-4} Tesla (T) / 1 Tesla = 10,000 Gauss
- Hull University Teaching Hospitals NHS Trust:

1.5T MR Scanners / 15,000G
3.0T MR Scanner / 30,000G

New £29.1 million national MRI scanning facility for Nottingham

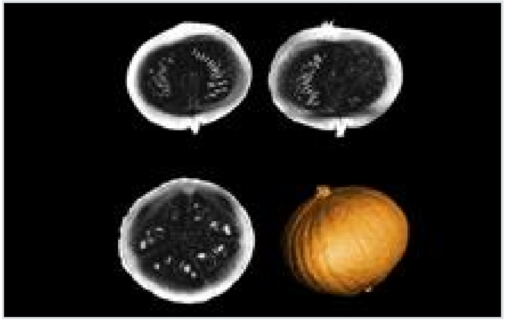
17 June 2022

The University of Nottingham has been awarded £29.1 million to establish the UK's most powerful Magnetic Resonance (MR) Imaging scanner as a national facility, subject to business case approval.

An 11.7T magnet with a 830-mm bore, which could be manufactured from niobium-titanium superconducting wire cooled to 2.2 K. The magnet will require passive shielding (~500 tonnes of iron).

WORLD PREMIERE

The most powerful MRI scanner in the world delivers its first images!



(c)CEA

September 2021, the 11.7 Tesla MRI of the Iseult project, the most powerful in the world for human imaging, has just unveiled its first images. They validate the entire process that has enabled, thanks to multiple technological breakthroughs, the transformation of an « outstanding » magnet, delivered in 2017 to the CEA-Paris-Saclay site, into an « imager ». This MRI, designed by CEA engineers and researchers together with Siemens Healthineers, will enable major advances in fundamental research, cognitive sciences and in understanding brain pathologies.

Published on 7 October 2021

Equilibrium magnetisation (M_0) and dependence on the strength of the magnetic field, B_0

Bay Area nurse crushed in MRI accident highlighting safety concerns

By Brooks Jarosz | Published October 27, 2023 6:34PM | Kaiser Permanente | KTVU FOX 2 |

Larmor Equation

- For a spin 1/2 nucleus in a magnetic field B_0, the difference between the energy levels is:

$$\Delta E = \gamma h B_0 / 2\pi \text{ (see Boltzmann Distribution)}$$

- A radio wave of frequency ω has energy given by:

$$E_\omega = h\omega/2\pi$$

- There using these two equations we can calculate the correct frequency to input radio waves at

- This is known as the Larmor Equation and ω_0 is the Larmor frequency:

$$\boldsymbol{\omega_0 = \gamma B_0}$$

γ is the gyromagnetic ratio (For 1H, γ=42.58 MHz/T)
h is Planck's constant (h = 6.63×10−34 J/Hz)

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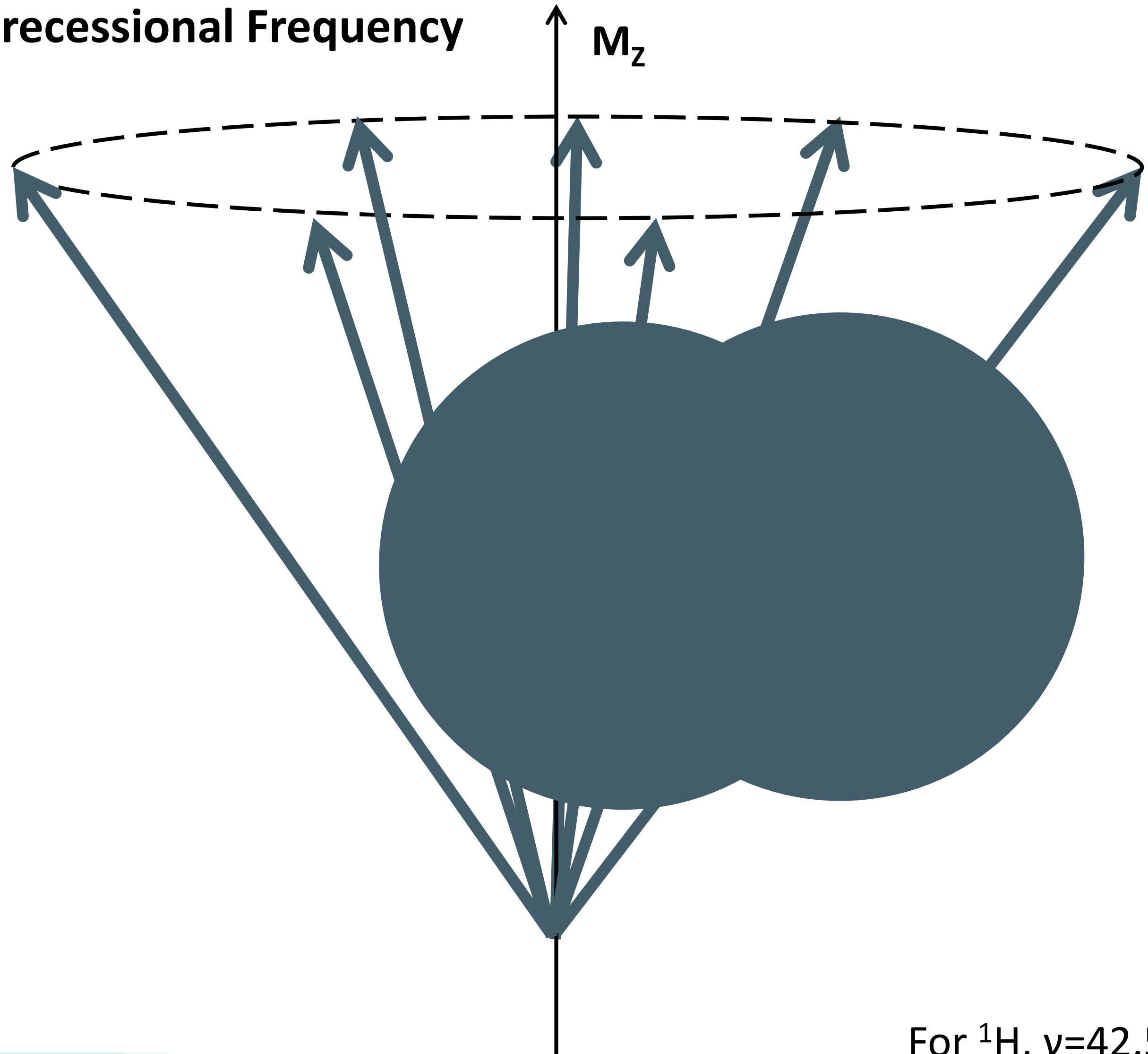
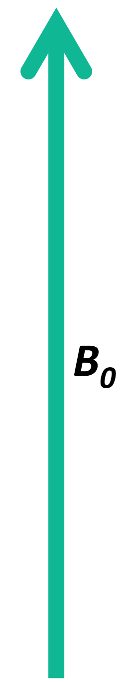
Precessional Frequency



$$\omega_0=\gamma B_0$$



For ^1H, γ=42.58 MHz/T (gyromagnetic ratio)

ω= angular frequency or Larmor frequency

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Larmor Equation

$$\omega=\gamma B_0$$

ω = precessional frequency or Larmor frequency
γ = gyromagnetic ratio (constant = 42.58 MHz/T)
B_0 = static magnetic field

- Precessional frequency increases as B_0 increases

$$\omega_{1.5T} = \gamma \times 1.5T = 63.87\ MHz$$

$$\omega_{3.0T} = \gamma \times 3.0T = 127.74\ MHz$$

- This is important for excitation! - B_1 RF pulses must be at the correct frequency for resonance to occur!

γ is the gyromagnetic ratio (For 1H, γ=42.58 MHz/T)
h is Planck's constant (h = 6.63×10−34 J/Hz)

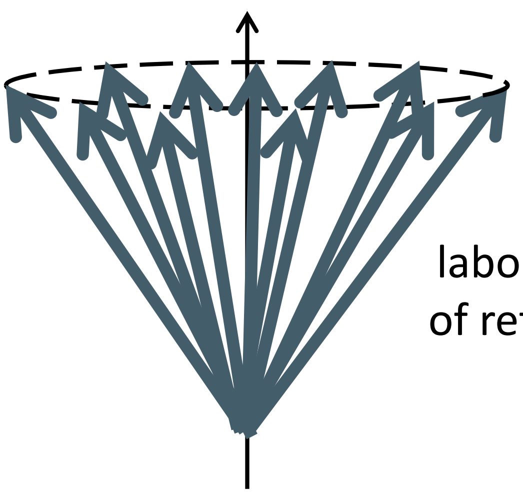
The Rotating Frame of Reference

- Using the Larmor Equation, we know that the spins are precessing at around 63.87 MHz in a 1.5T field

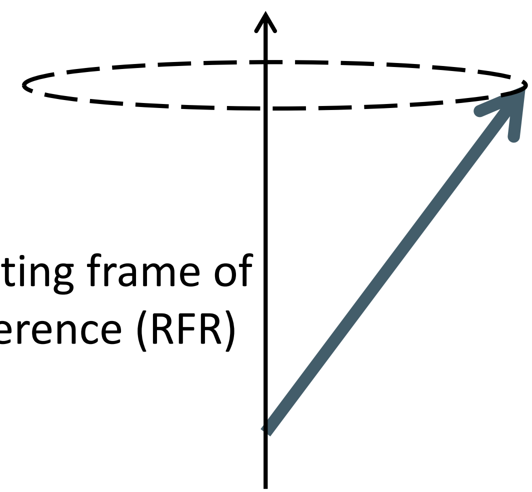
- e.g rotating 63.87 million times per second!

- Rather than visualise this, we observe the net magnetisation at the same rate.

- This is known as the rotating frame of reference



laboratory frame of reference (LFR)



rotating frame of reference (RFR)

Interaction with the Nucleus

- If we apply energy to the nucleus it can be absorbed and change the energy level of the nucleus

- Only happens if energy exactly matches the differences between the quantised levels.

Bohr Condition

- Energy is sent in as an electromagnetic wave with frequency ω (Radio wave)

- In MR, this is known as RF (**B_1**). $\omega_{1.5T} = \gamma \times 1.5T = 63.87$ MHz

- After a period of time, the nucleus returns to the lower energy level

- Energy is given back out and detected.

Excitation

- The plane in which in the net magnetisation exists can be manipulated using **RF pulses (B_1)**

- To stimulate the spins, an RF-coil must produce a time-varying excitation field ($B_1(t)$) with the following characteristics:
  - $B_1(t)$ must have components that rotate near the resonant frequency (ω_0)
  - $B_1(t)$ must have components perpendicular to the static magnetic field (B_0)

- For a simple RF pulse, a generator can be switched on and off, with the flip angle produced expressed as:

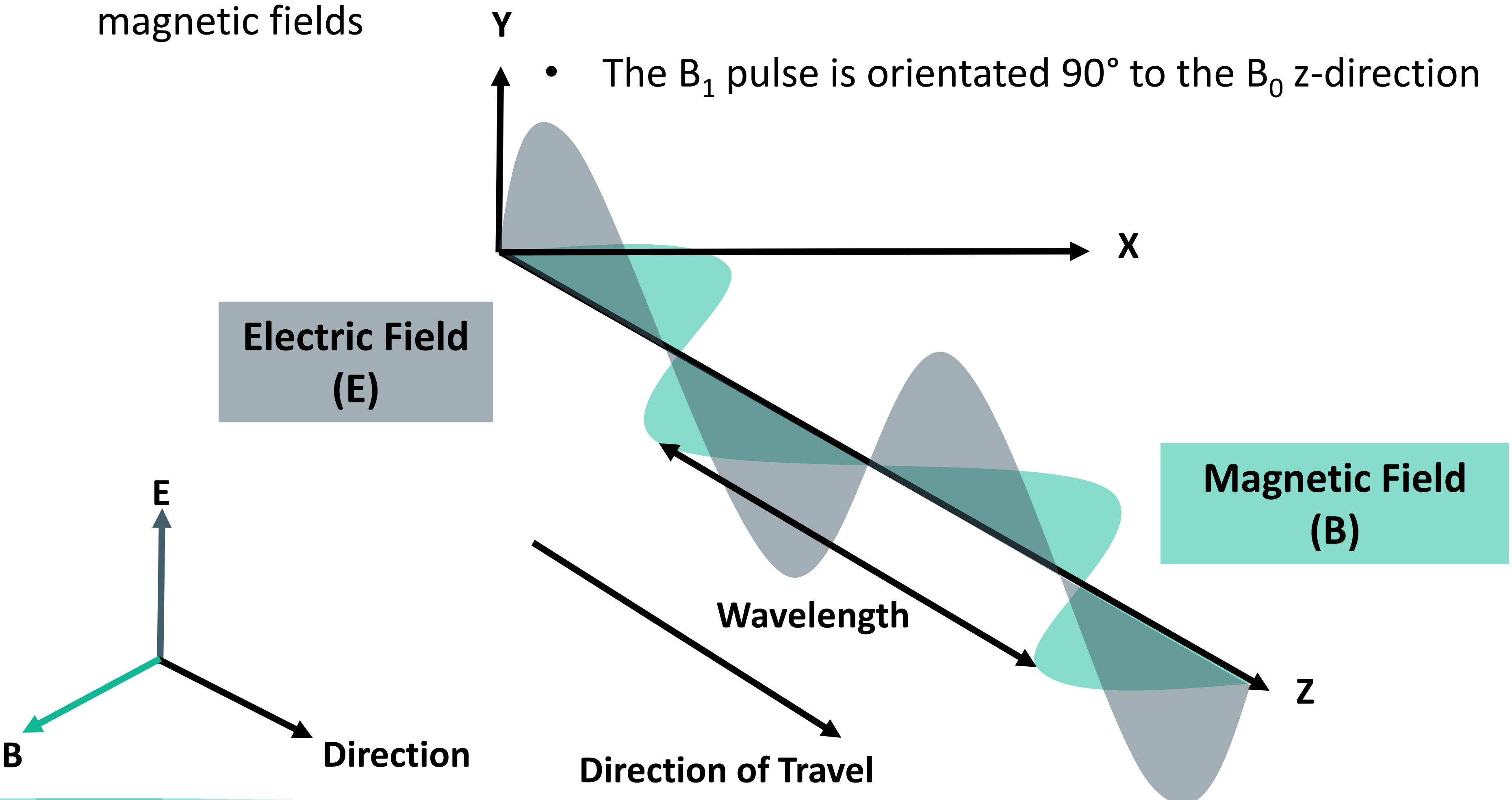
$$\alpha = \gamma B_1 t_p$$

- where α is the flip angle, γ is the gyromagnetic ratio, B_1 is the strength of the RF pulse and t_p is the RF pulse duration

Precession about magnetic fields (B_0 and B_1)

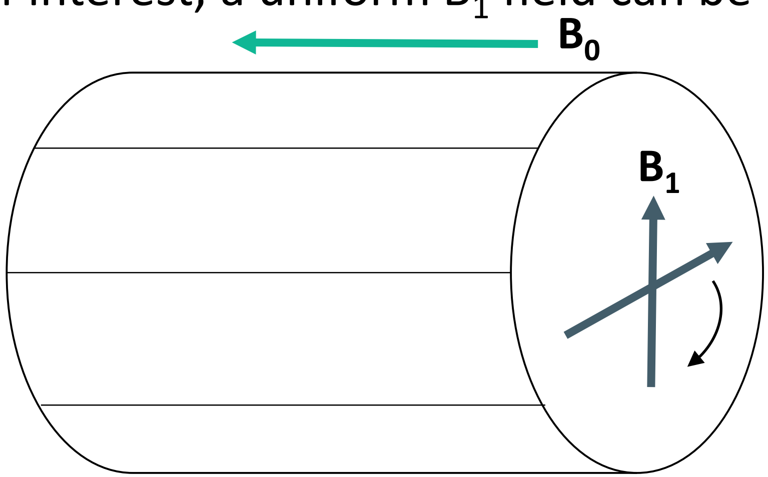
Radiofrequency magnetic field (B_1)

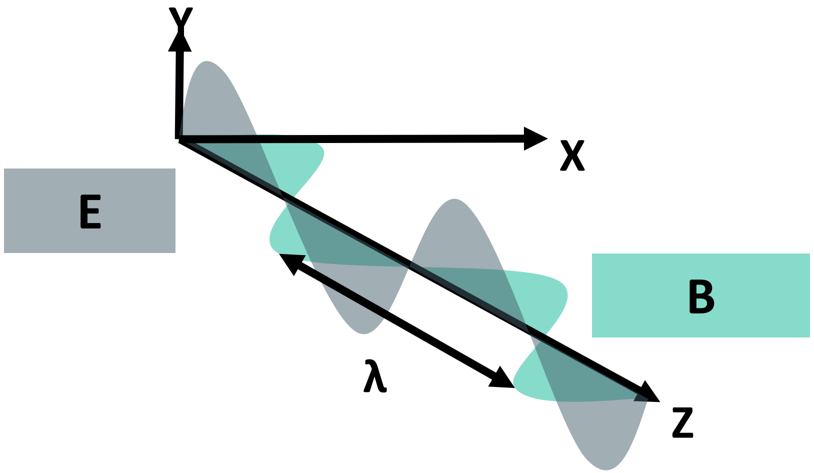
- RF pulses in MRI refer to electromagnet waves with orthogonal eclectic and magnetic fields
  - The B_1 pulse is orientated 90° to the B_0 z-direction



RF Transmission (Tx)

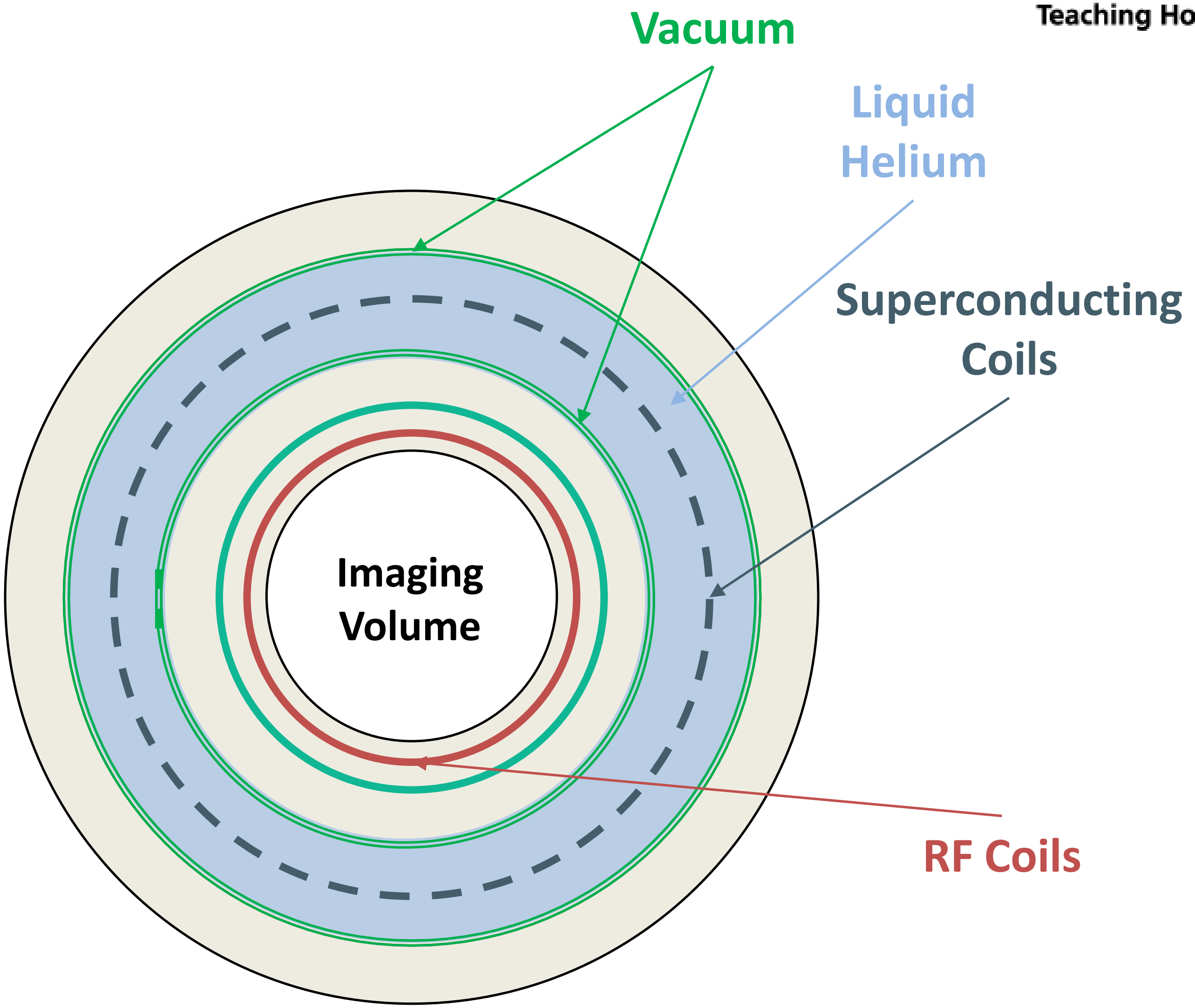
- For most examinations, the integral body RF coil is used for excitation
- Typically this a birdcage design which operates in quadrature to produce a circularly-polarised field
- This coil is incorporated into the main scanner
- By having a cylindrical design which incorporates the entire anatomical region of interest, a uniform B_1 field can be produced.





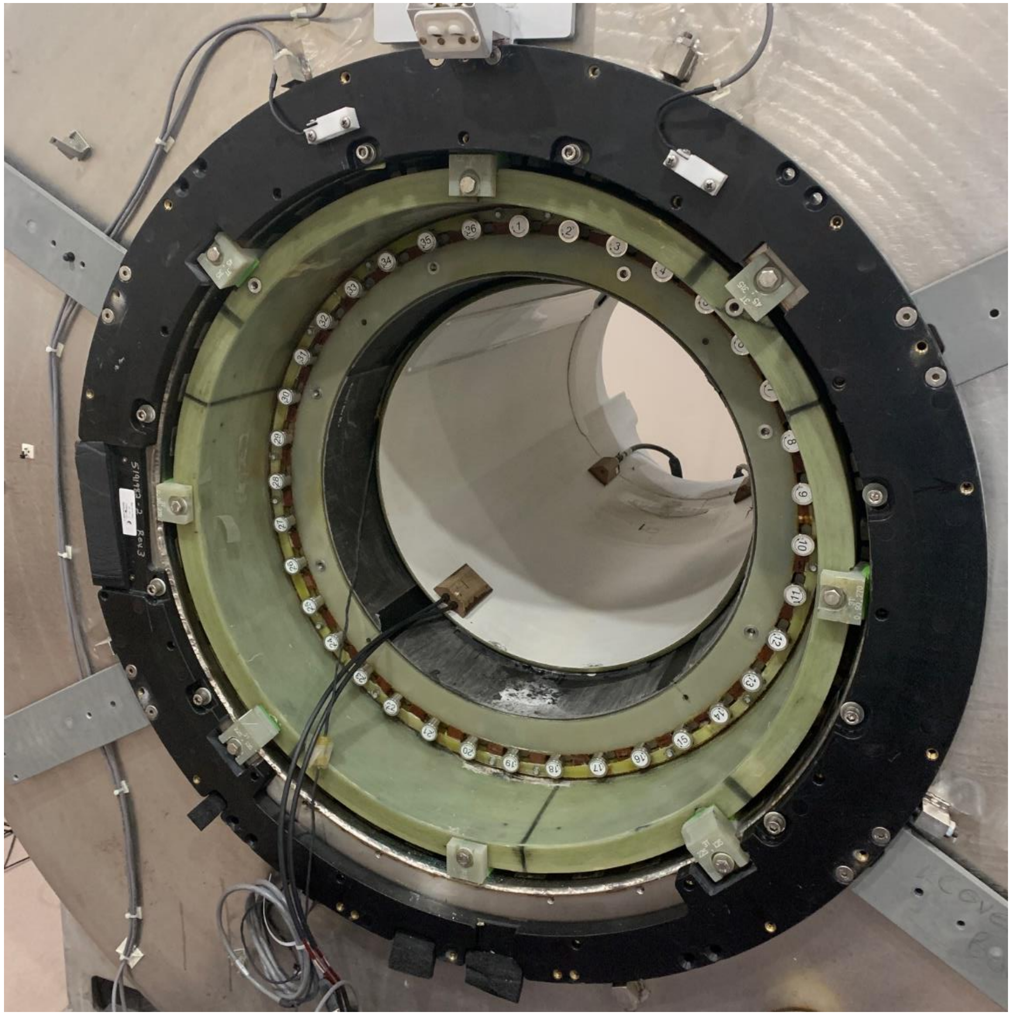
Precession about magnetic fields (B_0 and B_1)

System Design



Precession about magnetic fields (B_0 and B_1)

System Design



RF Receive Coil (Rx)

- The function of the receiver coil is to maximise signal detection, whilst minimising the noise
- Usually the major source of noise is from the patient's tissue (Brownian motion of electrolytes)
- To minimise the noise, and maximise the SNR, it is necessary to minimise the coil dimensions, i.e. the coil's volume should be filled as much as possible by the sample.
- There are two fundamental types of receiver coil: volume & surface.
  - **Volume coils** completely encompass the anatomy of interest and are often combined transmit/ receive
  - **Surface coils** are generally receive only, due to their inhomogeneous reception field

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Volume coils

- Radiofrequency coil that surrounds either the whole body, or one specific region, such as the head or a knee
- Volume coils have a better RF homogeneity than surface coils, which extends over a large area
- The most commonly used design is a (birdcage) bird cage coil. This consists of a number of wires running along the z-direction
- It is possible to use the same coil to transmit and receive (Tx/Rx, T/R), or to use two separate coils (Tx only, Rx only)

T/R Coils Available:

- **GE Artist/Premier:** Integral Body; Knee; Hand-Wrist; Quadrature Head (Birdcage)
- **Siemens Sola/Aera:** Integral Body; Knee

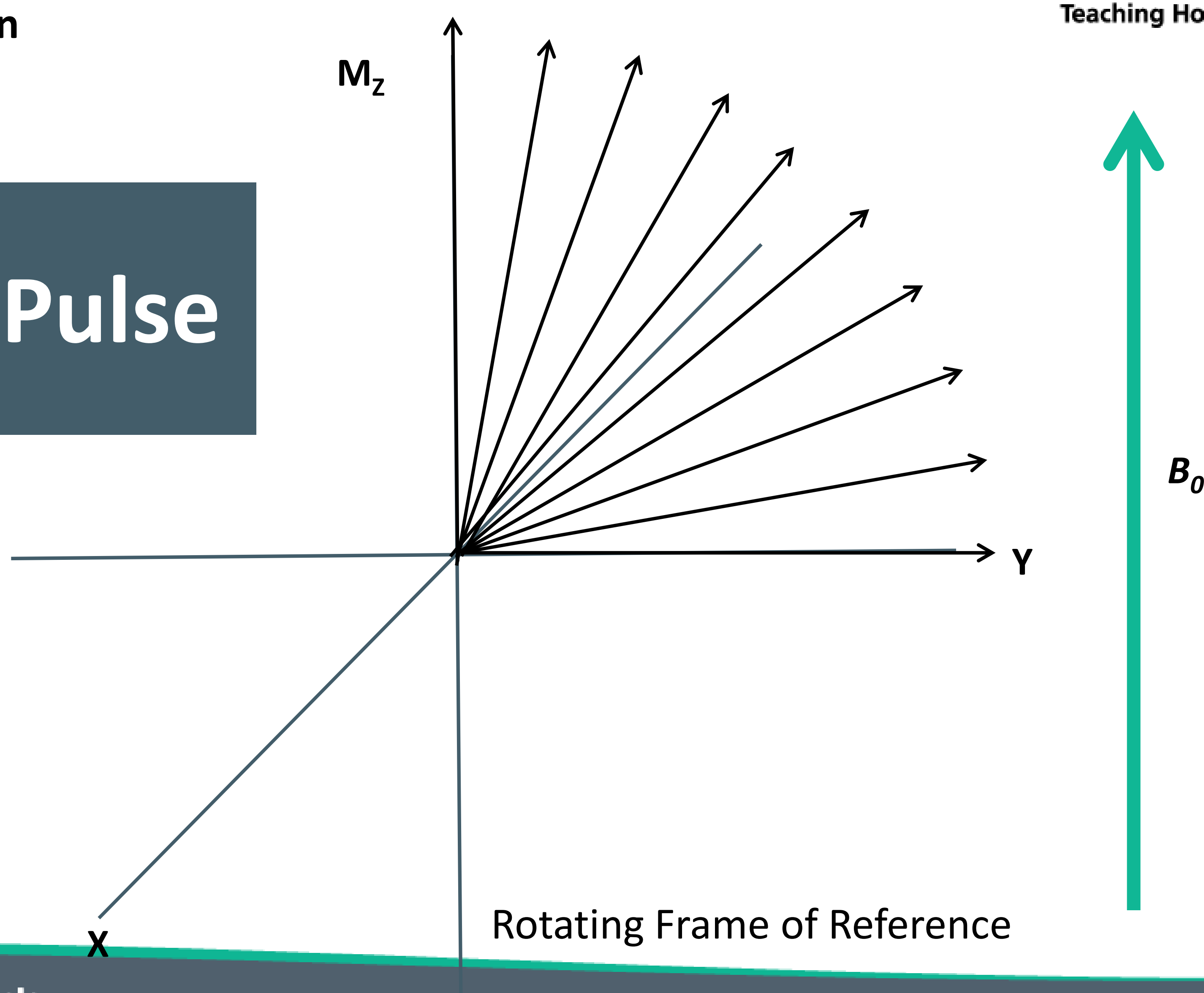
T_1 Relaxation (spin-lattice relaxation)

- Following a B_1 excitation pulse (RF), longitudinal magnetisation begins to recover immediately
- *Spin-lattice relaxation* is the term describing the release of energy back to the *lattice* (the molecular arrangement and structure of the hydration layer), and the regrowth of M_z
- T_1 is the time needed for the recovery of 63% of M_Z after a 90-degree pulse

Longitudinal (M_z) and transverse magnetisation (M_{xy})

T_1 Relaxation

90° Pulse



Rotating Frame of Reference

Longitudinal (M_z) and transverse magnetisation (M_{xy})

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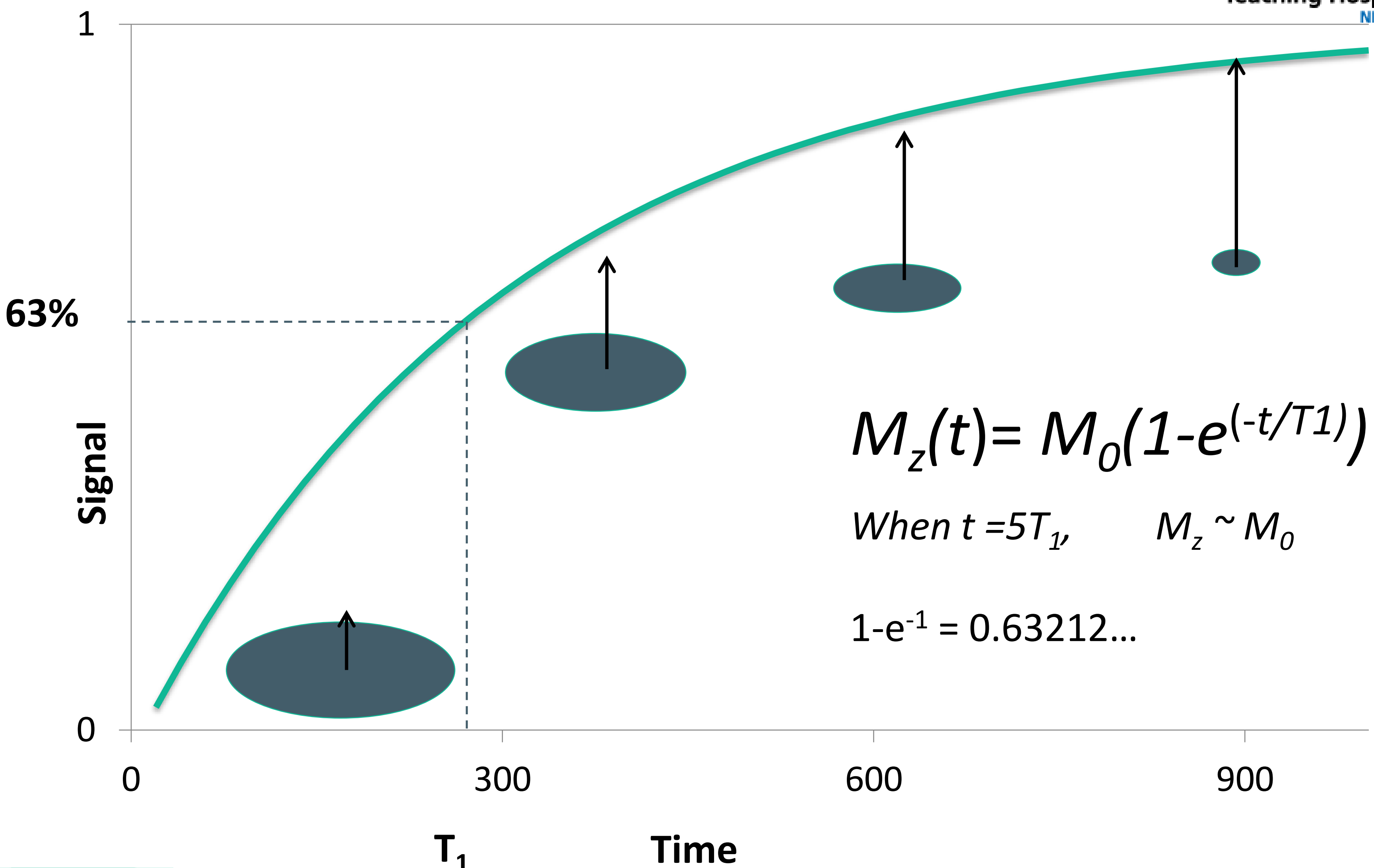


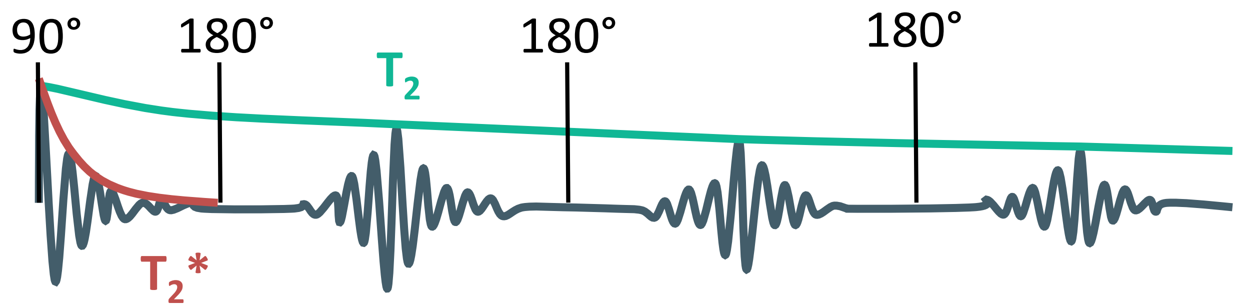
$$M_z(t) = M_0(1-e^{(-t/T1)})$$

When $t = 5T_1$, $M_z \sim M_0$

$1-e^{-1} = 0.63212...$

T_2 Relaxation (spin-spin relaxation)

- T_2 decay is the process whereby spins begin to dephase, occurring **simultaneously with T_1 relaxation**
- Due to individual spins observing local differences in the magnetic field **caused by interactions between spins**
- Spins dephase much quicker than the 'true' T_2 due to inhomogeneities in the static magnetic field (B_0) causing the signal decay to be characterised as T_2*



T_2/T_2* Relaxation

$$\frac{1}{T_2^*} = \frac{1}{T_2} + \frac{1}{T_2'}$$

T_2' represents:

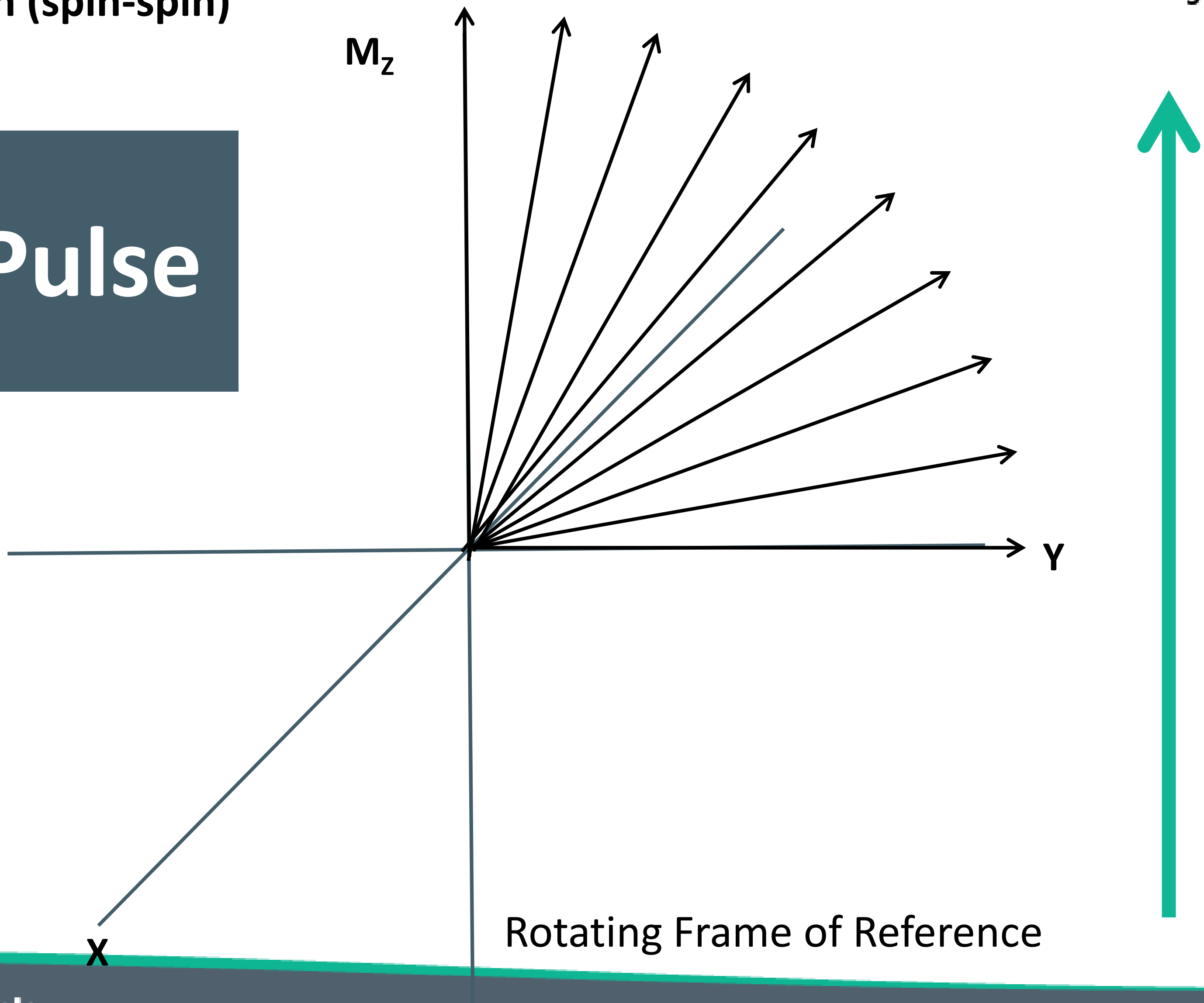
- imperfections in the field
- variations in B_0 field inhomogeneities & magnetic susceptibility
    - Susceptibility is a property of matter which determines how easily it becomes magnetised when placed in an external field
    - Susceptibility artefacts are signal voids present due to differences in susceptibility between objects and tissue or air/tissue interface

- T_2 - spin-spin relaxation which is tissue specific

Longitudinal (M_z) and transverse magnetisation (M_{xy})

T_2 Relaxation (spin-spin)

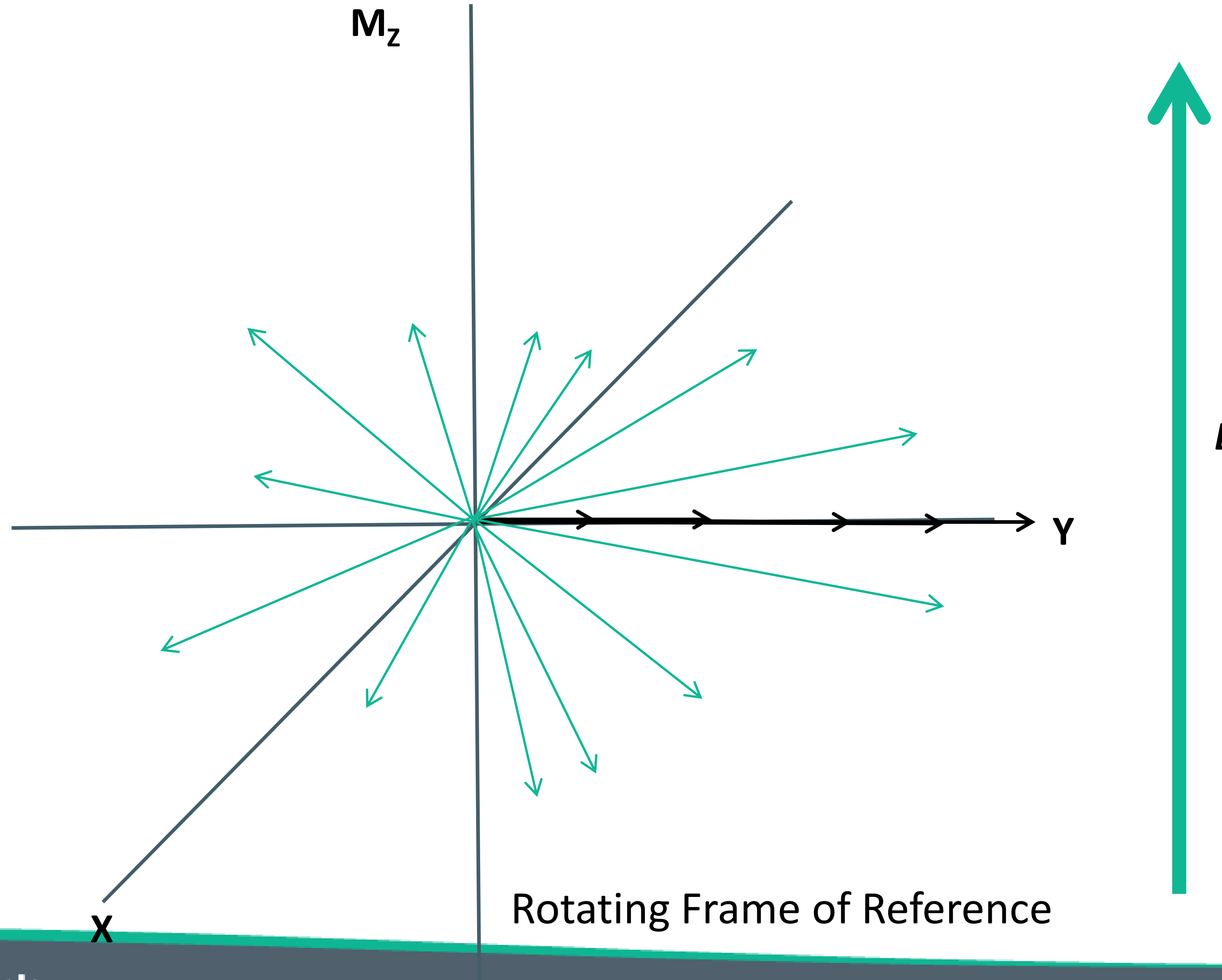
90° Pulse



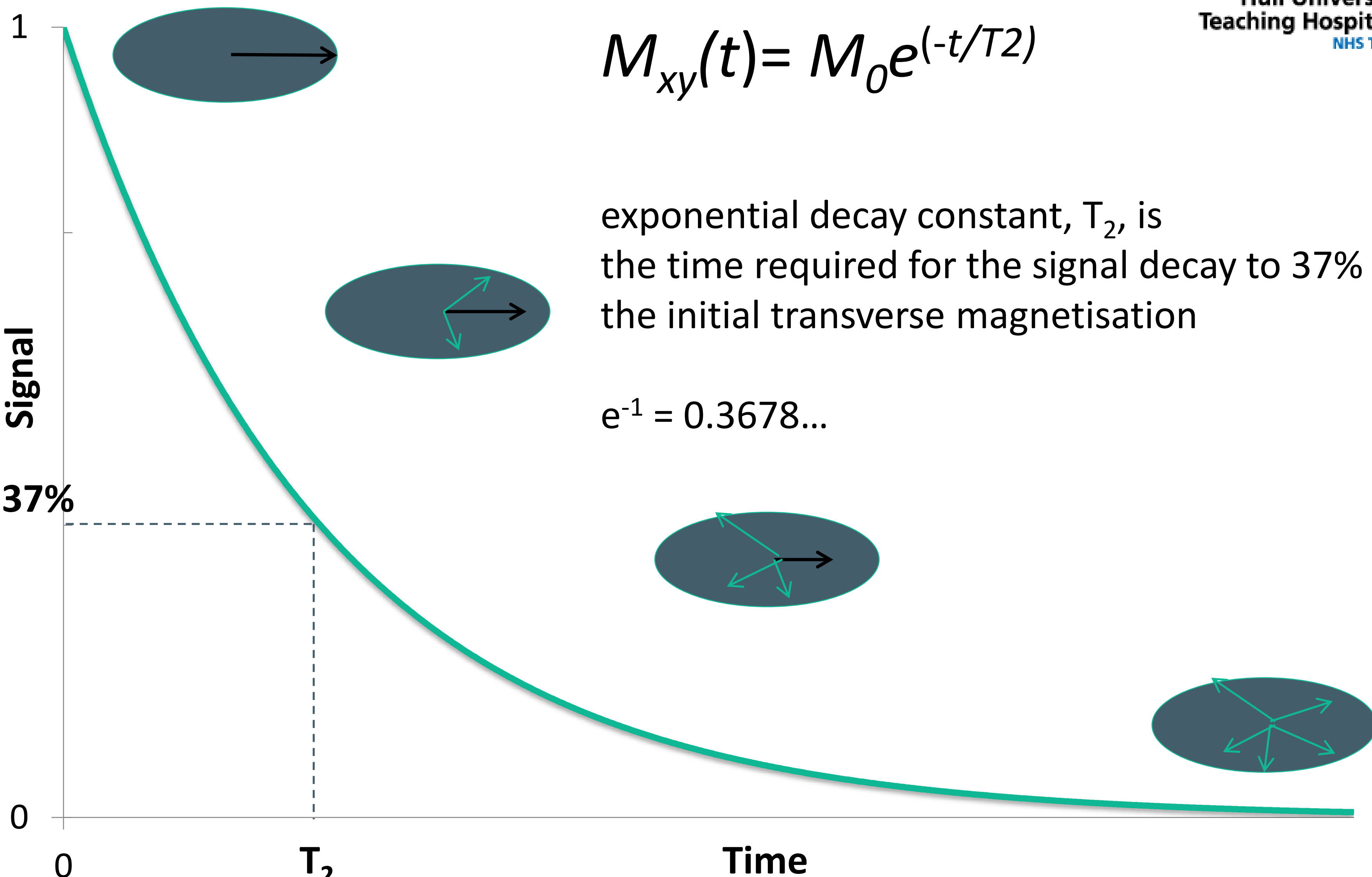
Rotating Frame of Reference

Longitudinal (M_z) and transverse magnetisation (M_{xy})

T_2 Relaxation (spin-spin)



Longitudinal (M_z) and transverse magnetisation (M_{xy})



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

exponential decay constant, T_2, is the time required for the signal decay to 37% of the initial transverse magnetisation

$e^{-1} = 0.3678…$

Longitudinal (M_z) and transverse magnetisation (M_{xy})

PHYSICAL REVIEW VOLUME 70, NUMBERS 7 AND 8 OCTOBER 1 AND 15, 1946

Nuclear Induction

F. Bloch
Stanford University, California
(Received July 19, 1946)

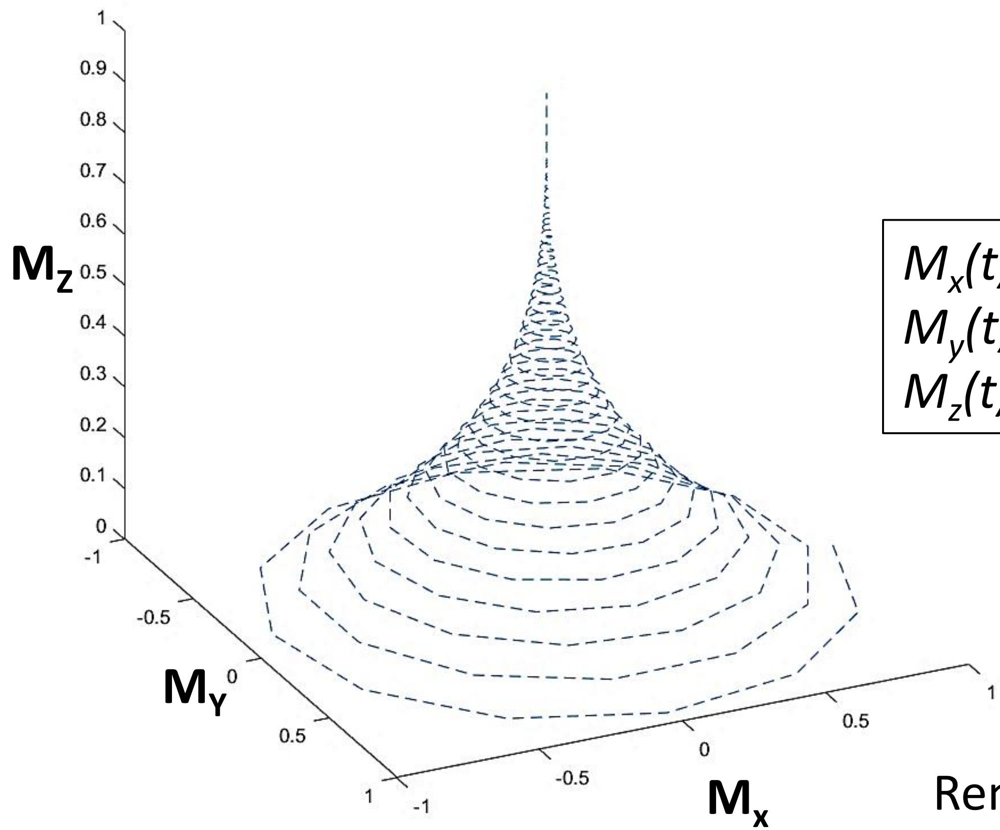
$$M_x(t) = M_o\, e^{-t/T2} \sin \omega t$$
$$M_y(t) = M_o\, e^{-t/T2} \cos \omega t$$
$$M_z(t) = M_o\, (1 - e^{-t/T1})$$

The magnetic moments of nuclei in normal matter will result in a nuclear paramagnetic polarization upon establishment of equilibrium in a constant magnetic field. It is shown that a radiofrequency field at right angles to the constant field causes a forced precession of the total polarization around the constant field with decreasing latitude as the Larmor frequency approaches adiabatically the frequency of the r-f field. Thus there results a component of the nuclear polarization at right angles to both the constant and the r-f field and it is shown that under normal laboratory conditions this component can induce observable voltages. In Section 3 we discuss this nuclear induction, considering the effect of external fields only, while in Section 4 those modifications are described which originate from internal fields and finite relaxation times.

- Bloch equations describe behaviour of the magnetic moment of a sample of weakly interacting spins

Free Induction Decay (FID)

- Bloch equations describe behaviour of the magnetic moment of a sample of weakly interacting spins



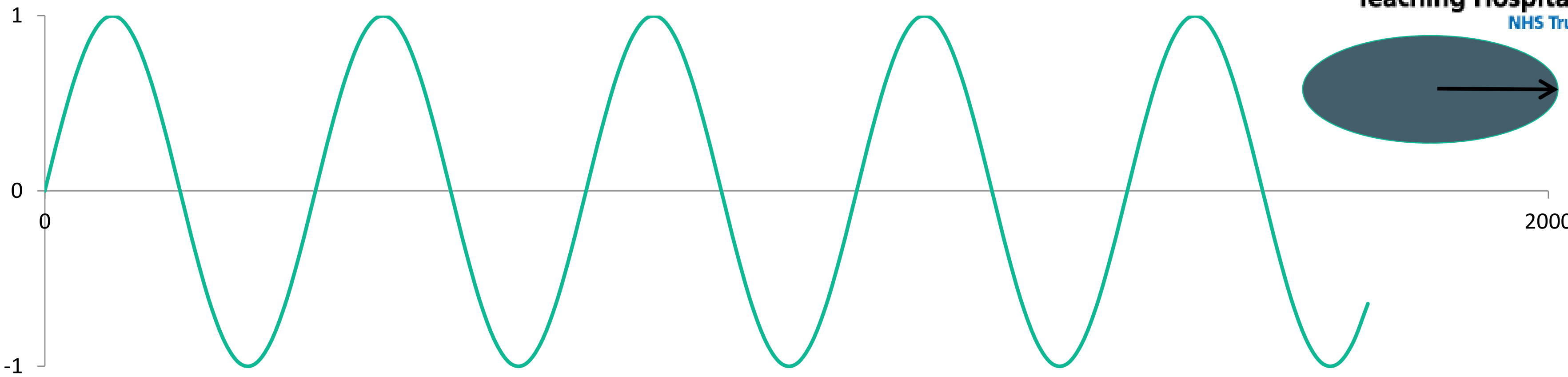
$$M_x(t) = M_o\, e^{-t/T2} \sin \omega t$$
$$M_y(t) = M_o\, e^{-t/T2} \cos \omega t$$
$$M_z(t) = M_o\, (1 - e^{-t/T1})$$

Remember! - T_2 decay occurs **simultaneously with T_1 relaxation**

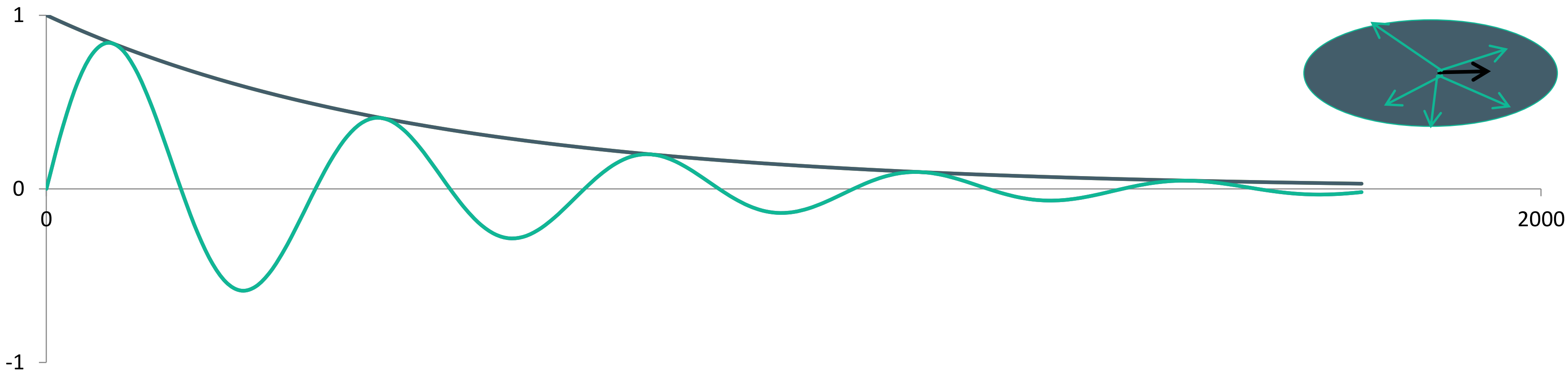
Free Induction Decay (FID)

- In MR, free induction decay is the observable NMR signal generated by non-equilibrium nuclear spin magnetization precessing about the magnetic field (typically along z)
- Receiver coil sees oscillating magnetic field which induces a varying voltage
- Coils only measures signal in transverse (M_{xy}) plane
- Relaxation occurs due to interactions between spin-lattice (T_1) and spin-spin (T_2)
- FID is attenuated by characteristic relaxation time T_2*

Longitudinal (M_z) and transverse magnetisation (M_{xy})



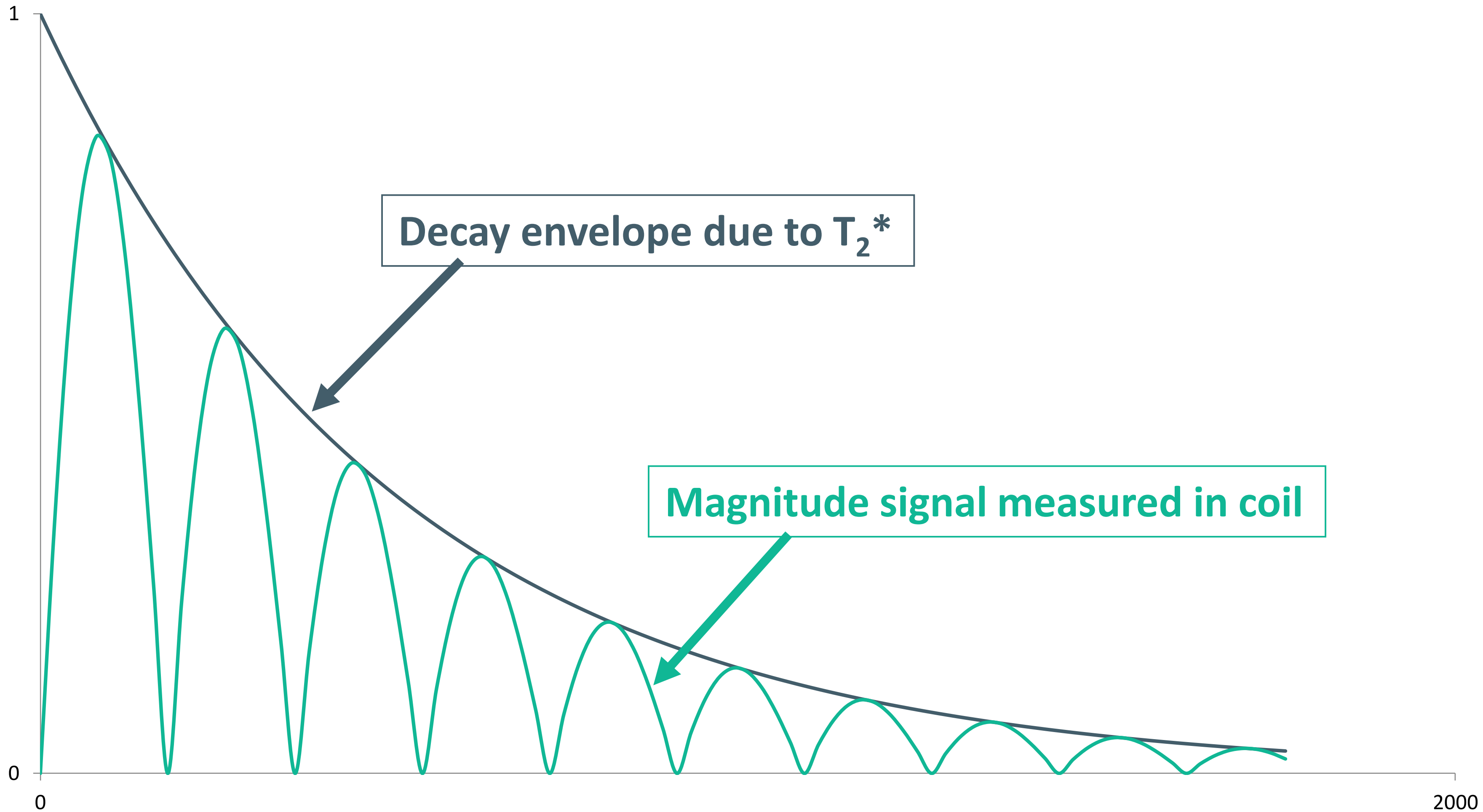
With no relaxation, the signal is sinusoidal



However, signal is attenuated (sinc function) due to relaxation (FID)

Longitudinal (M_z) and transverse magnetisation (M_{xy})

Free Induction Decay (FID)



Longitudinal (M_z) and transverse magnetisation (M_{xy})

Surface coils

- Placed close to the area under examination
- Consist of a single or double loop of wire
- High signal to noise ratio (SNR)
- However; signal uniformity falls off quickly away from the coil
- Limited Field of View (FOV)

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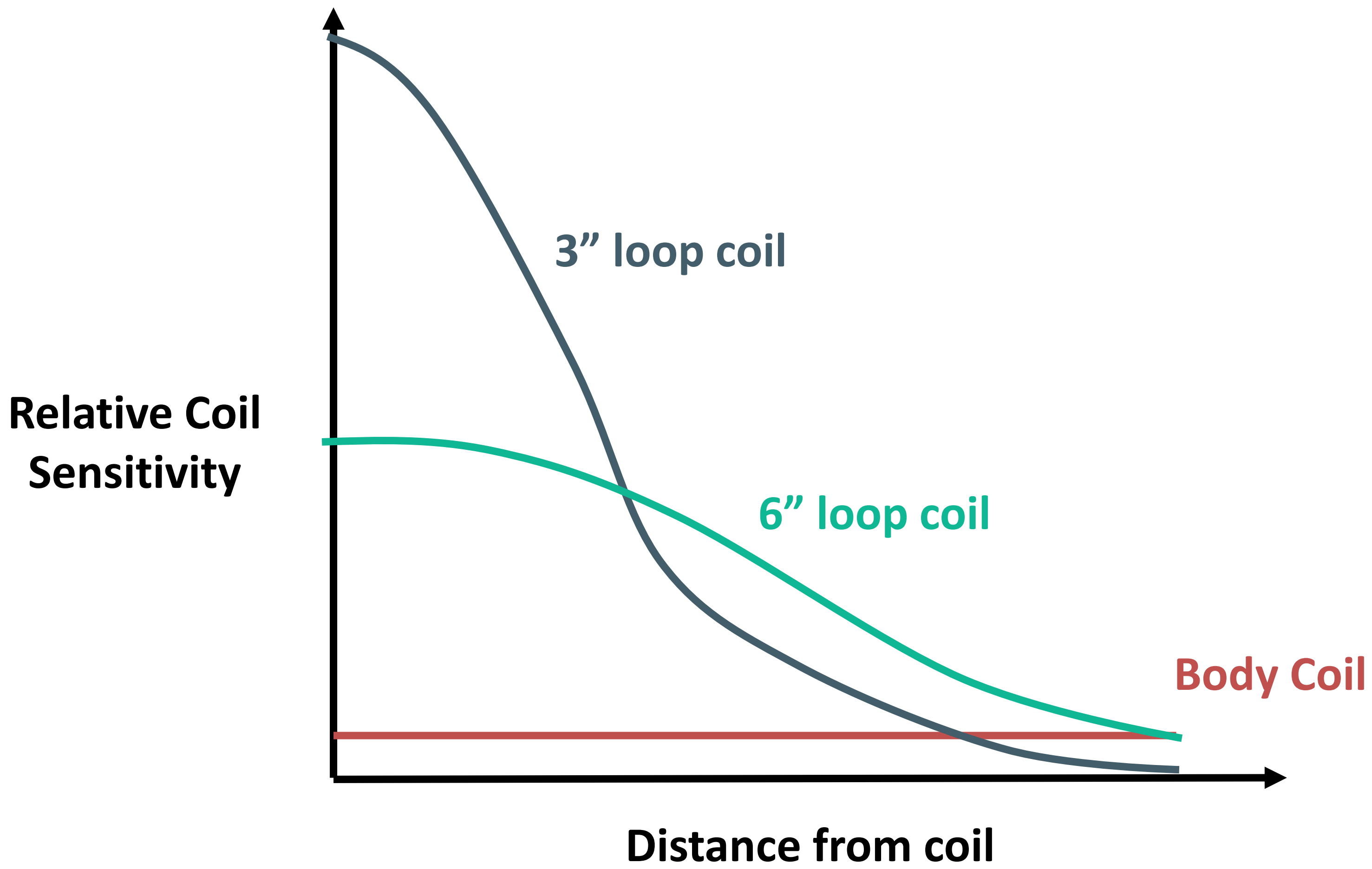
RF Coils: Signal Characteristics



Phased Array Coils

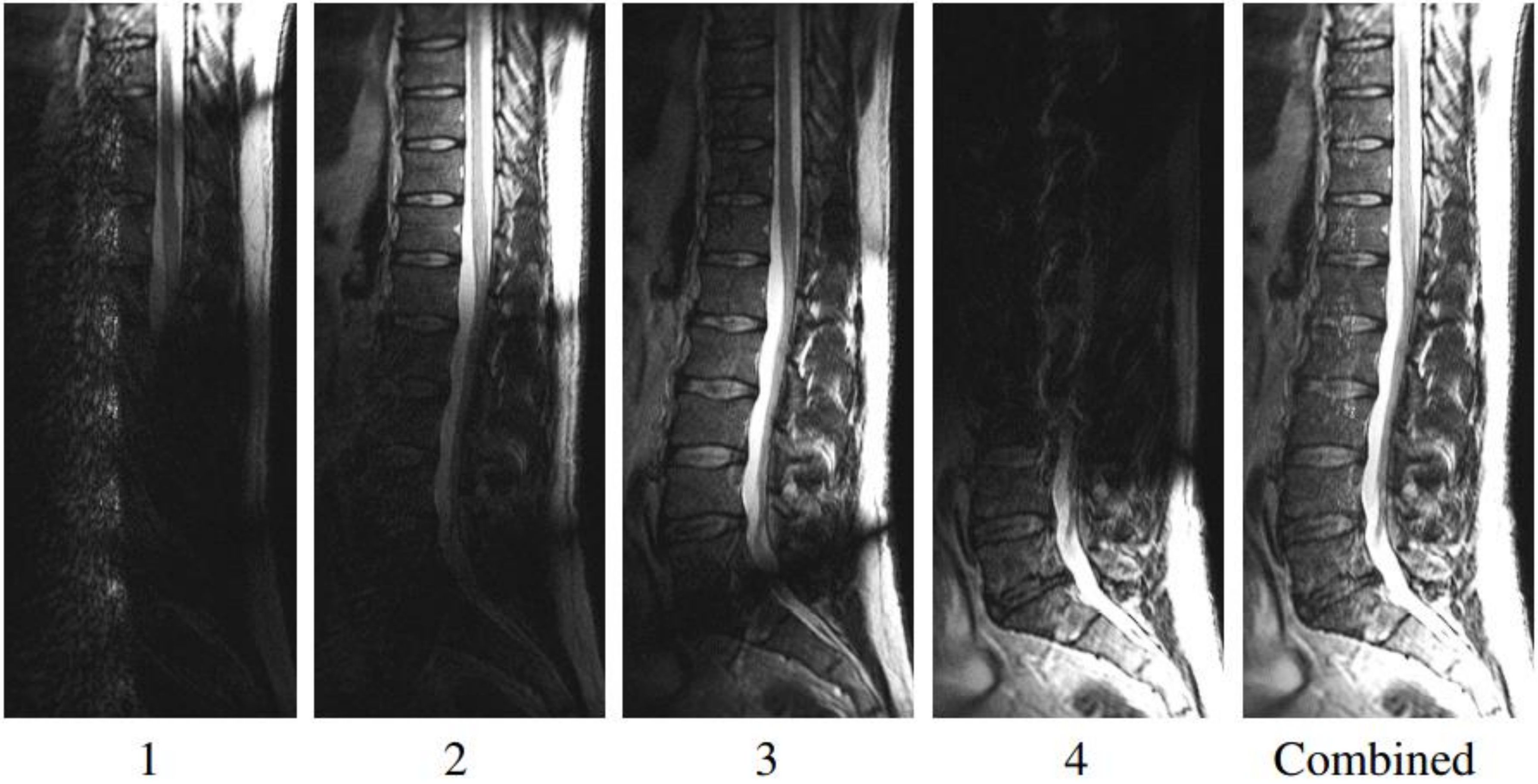
- The problem of the limited FOV of surface coils can be solved by using a number of coils simultaneously, known as an array
- Array coils are now available to image most areas, e.g. a spine array coil or a torso array coil.
- These coils need to be constructed very carefully so that the individual coils (also known as elements) do not interact with each other, a phenomenon known as coupling which reduces SNR in the images
- One way of effectively 'decoupling' one element from its neighbours is to geometrically overlap the coils in a particular way
- Each element is connected to an entirely separate preamplifier and receiver which has the advantage that the noise in each receiver is completely different, i.e. uncorrelated, resulting in a higher SNR in the final image

Phased Array Coils

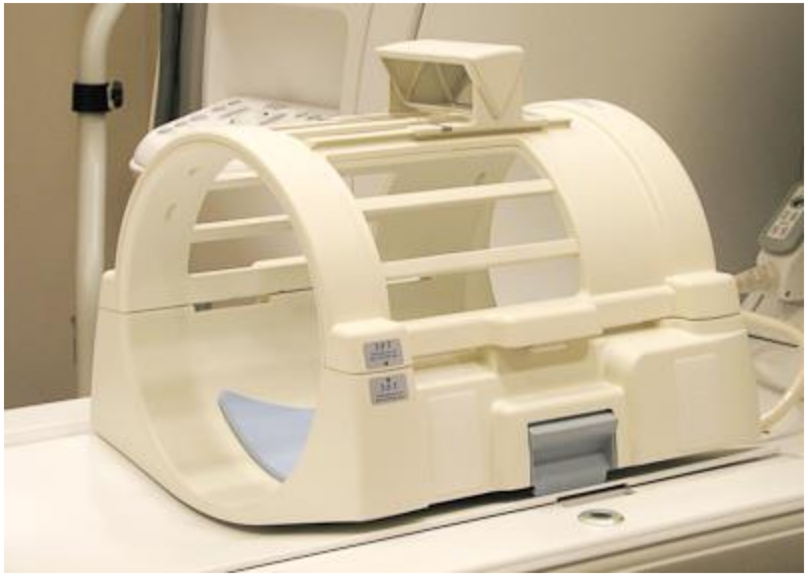
- Images can be produced from the individual elements, but generally they are combined together to produce an image with the large FOV advantage of the array but with the SNR advantage of the small individual elements
- Most common method for combining the signals is the ‘sum-of-squares’ technique, where the signal in a voxel of the final image is the sum of the squared signals intensities for that voxel measured by each element of the coil
- Phased array coil imaging generates more data than single receive coils, and reconstruction times may be longer due to the additional processing required
- A key advantage of multi-element coils is the possibility of using parallel imaging techniques for acceleration and reduced scan times

Longitudinal (M_z) and transverse magnetisation (M_{xy})

Phased Array Coils

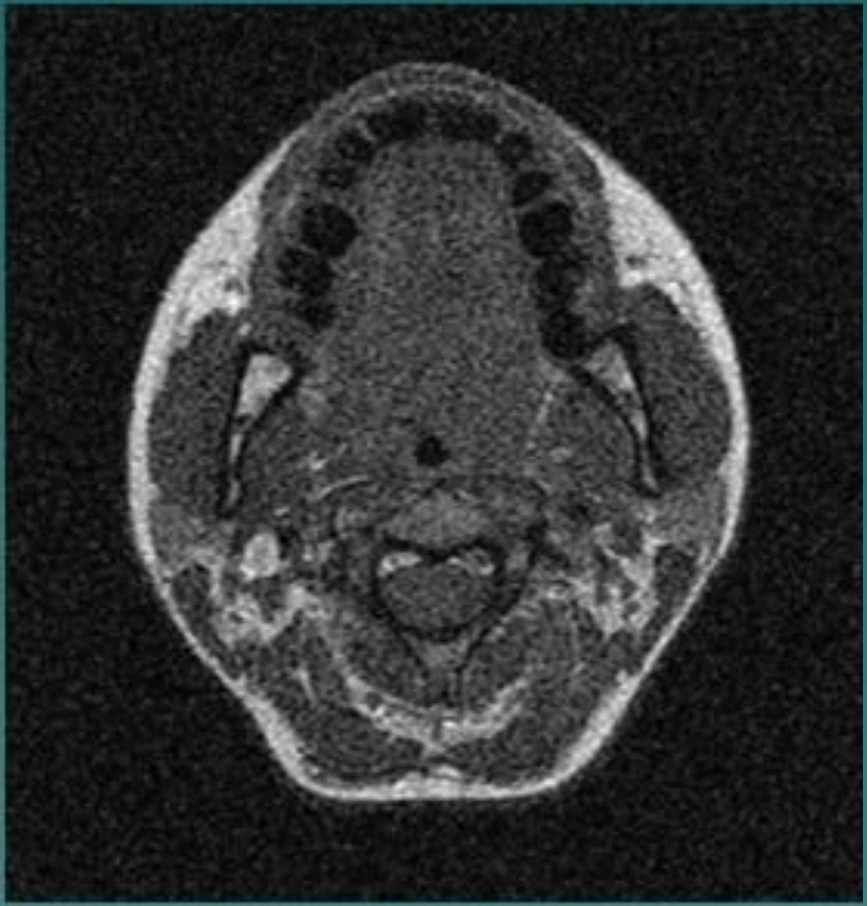


RF Coils

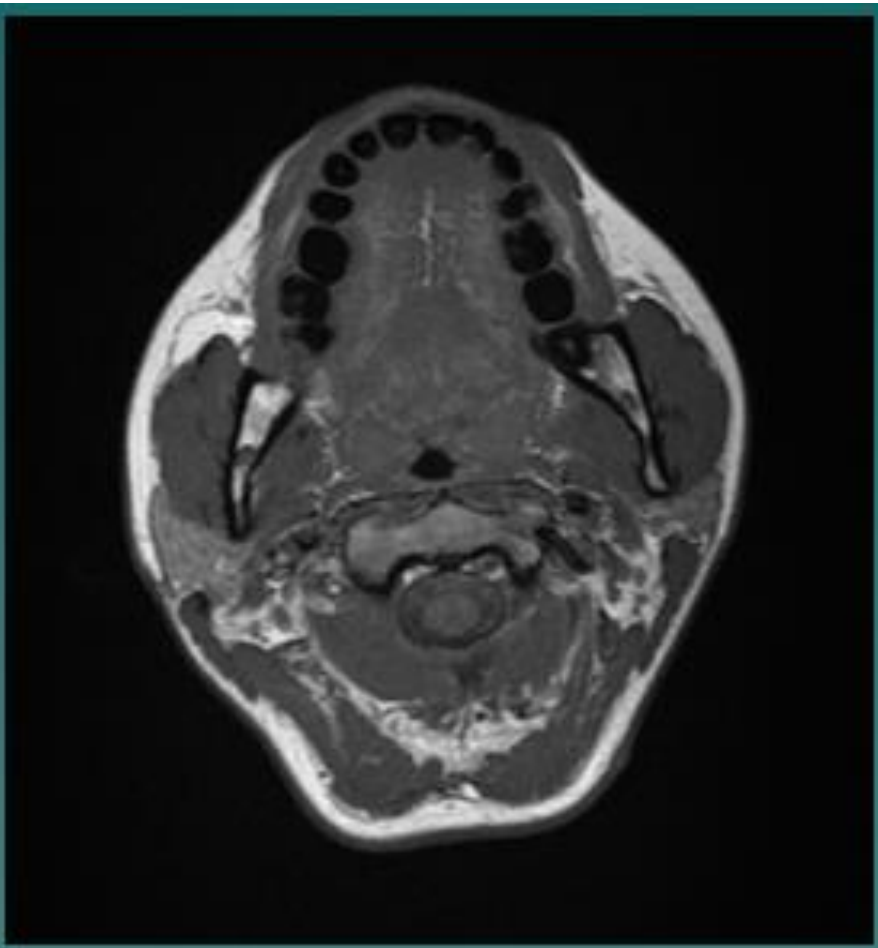


T/R Head Coil	**Phased Array Head Coil**
Local RF Transmission	Body Coil RF Transmission
Volume Coil Design	Phased Array Design
1 Receive Channel	48 Receive Channels
No Parallel Imaging	Parallel Imaging
Uniform signal across imaging volume	Increased signal close to the coils

RF Coils: Signal Characteristics



Body Coil Receive rSNR=1



HNS Coil Receive rSNR=25

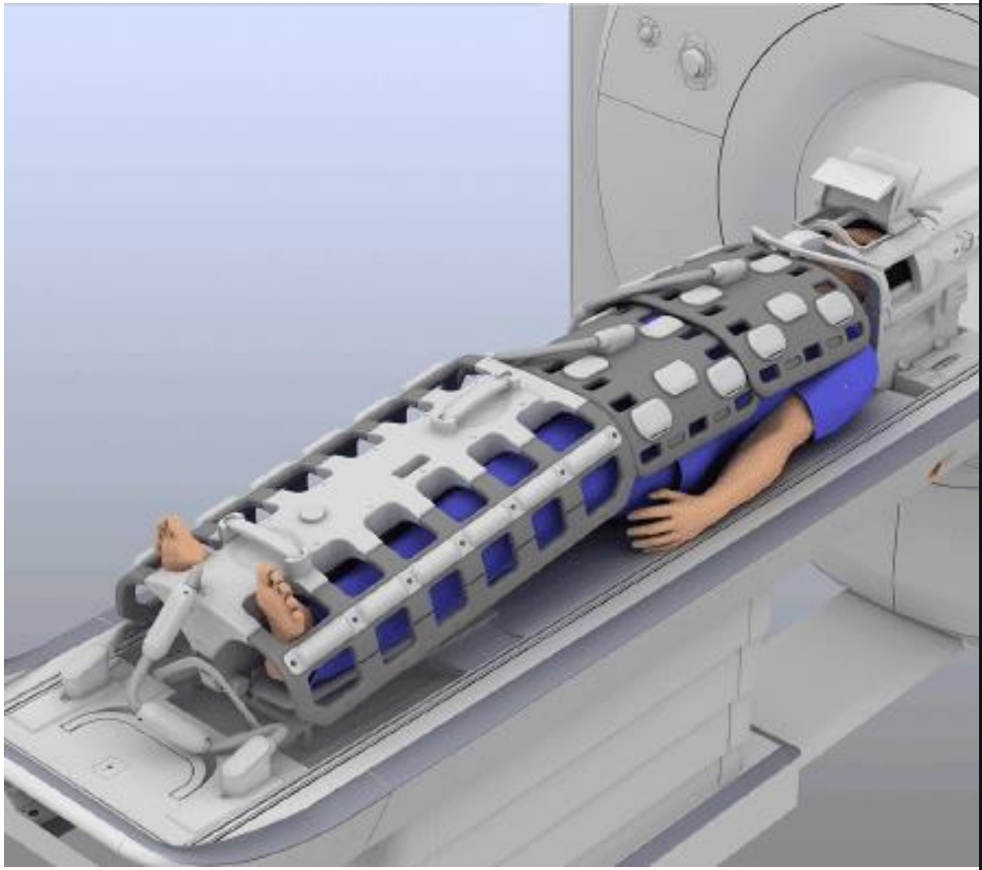
Longitudinal (M_z) and transverse magnetisation (M_{xy})

- 8-64 element coils are common now
- 64, 128 and even higher channel systems are now available (200+)
- Connecting two or more array coils simultaneously provides the possibility of whole body screening where the patient is imaged from head to toe using several stations
- With this technique, a particular FOV is scanned (e.g. the head and neck), then the patient is automatically moved into the magnet by a fixed distance and the next station is scanned (e.g. the upper thorax). This is repeated multiples times to provide complete coverage without having to change coils

Dedicated Scanner Coils

- Head/Neck
- Body
- Breast
- Wrist
- Shoulder
- Knee
- Flex (S/L)
- Per

Longitudinal (M_z) and transverse magnetisation (M_{xy})



Spin echo (SE) pulse sequences

- Collection of the Free Induction Decay (FID) signal is problematic as simultaneously transmitting and receiving RF can damage the scanner
- Spin Echo (SE) describes a simple pulse sequence capable of producing an "echo" later time when no RF is being transmitted
- The permits data collection and the creation of an MR image

Spin echo works as follows:

- Spin echo describes the excitation of the magnetised protons in a sample with a **90° RF pulse** and production of a FID, followed by a refocusing **180° RF pulse** to produce an echo
- The 90° pulse converts M_z into M_{xy} and creates coherent transverse magnetisation that immediately begins to decay at a rate described by T_2* relaxation

Longitudinal (M_z) and transverse magnetisation (M_{xy})

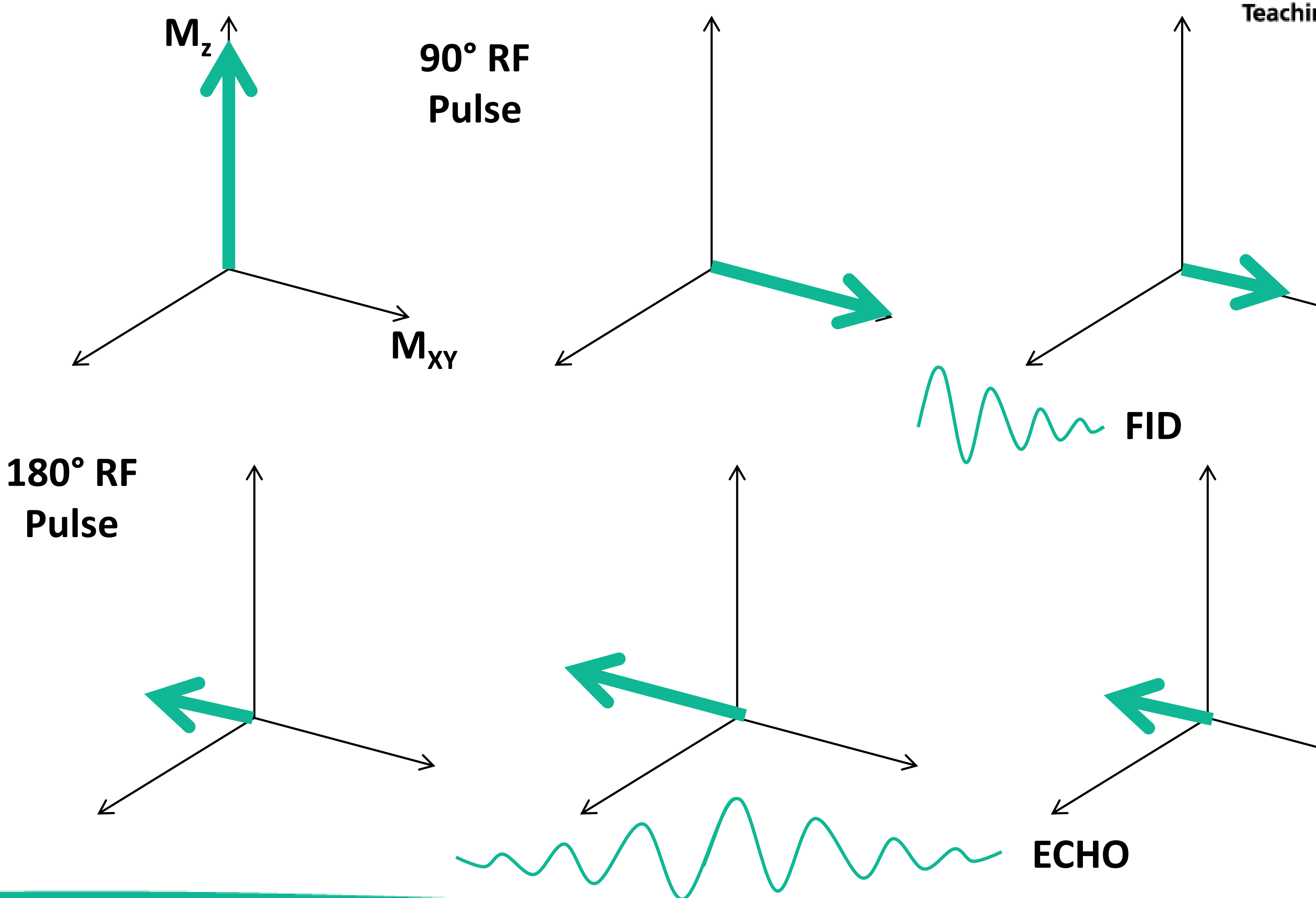
- The 180° RF pulse applied at TE/2 inverts the spins and induces phase coherence at TE

- Inversion of the spins causes the protons to experience external magnetic field variations opposite of that prior to TE/2, resulting in the cancellation of the extrinsic inhomogeneities and associated dephasing effects

- Subsequent 180° RF pulses during the TR interval produce corresponding echoes with peak amplitudes that are reduced by intrinsic T_2 decay of the tissues, and are immune from extrinsic inhomogeneities

- Digital sampling and acquisition of the signal occurs in a time window symmetric about TE, during the evolution and decay of each echo

Longitudinal (M_z) and transverse magnetisation (M_{xy})



Longitudinal (M_z) and transverse magnetisation (M_{xy})

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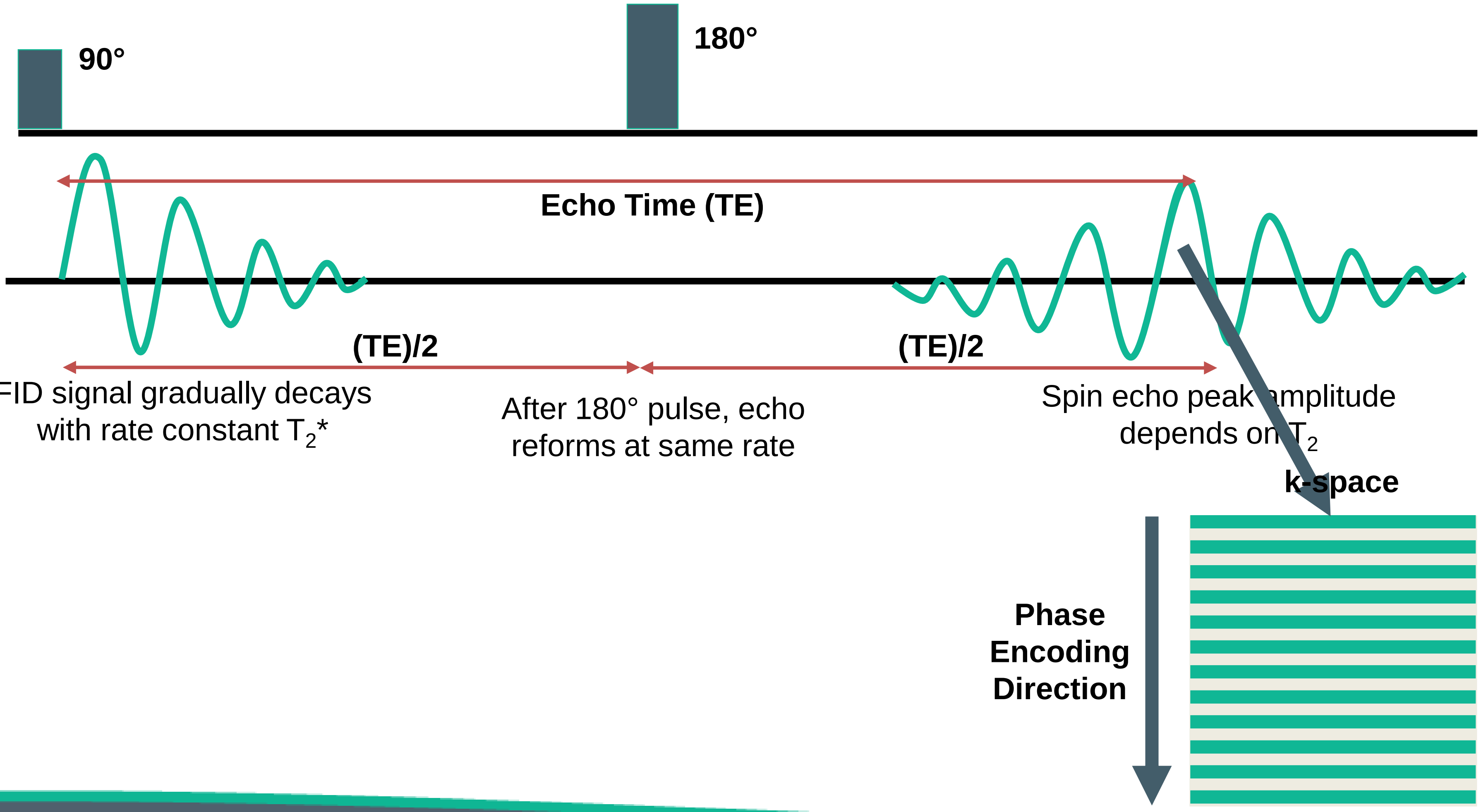
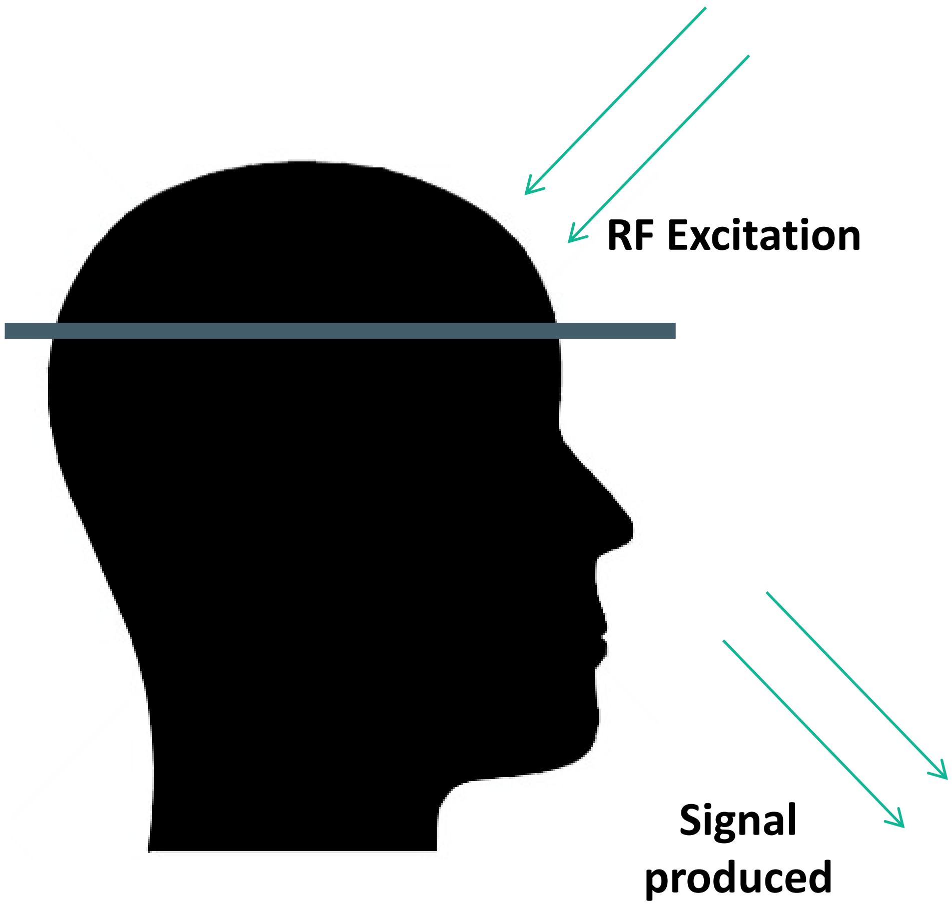
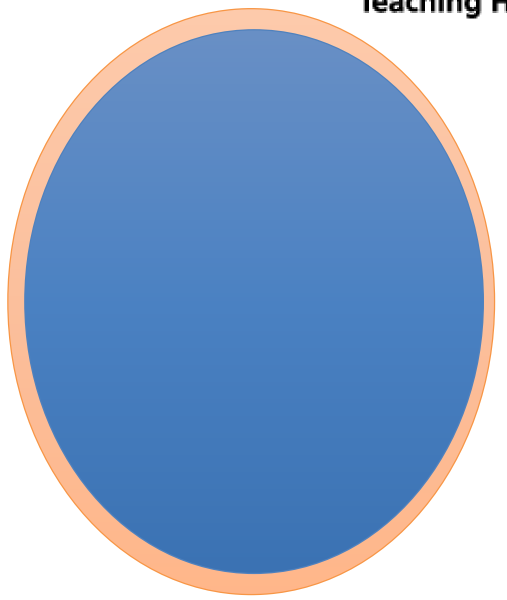
- Spin Echo (SE)
- Simple pulse sequence. Repeated multiple times to collect all the data

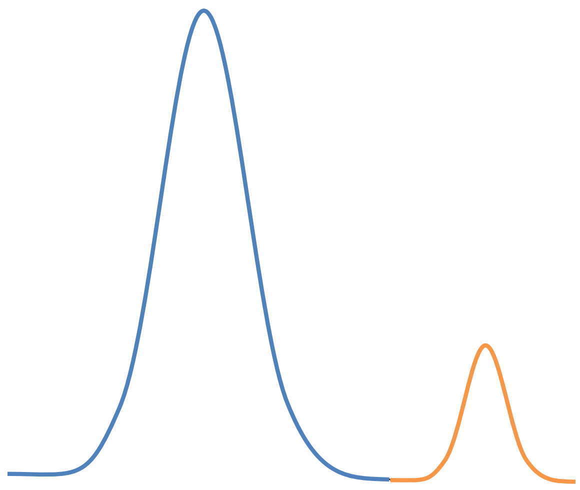


Slice Selection

Localisation







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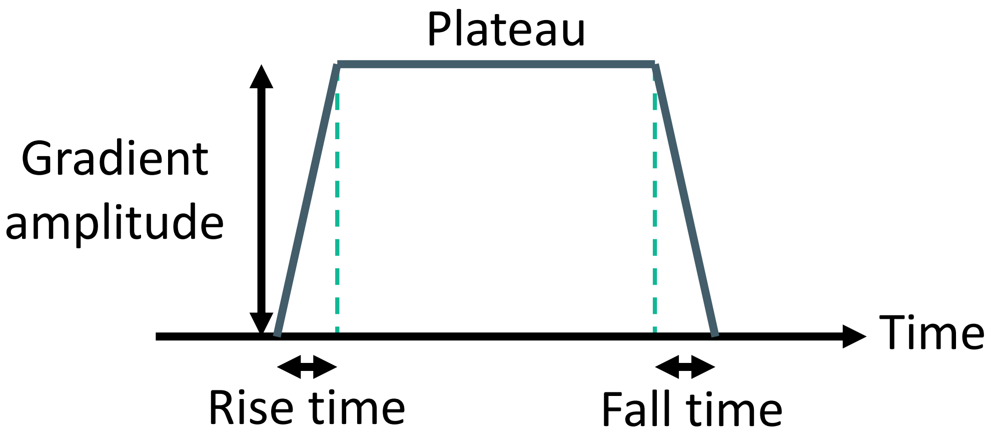
Introduction

- MR signal localisation utilises **magnetic field gradients**
- 2D slices are produced by the combination of an excitation RF pulse and **simultaneous slice-select gradient**
- the in-plane MR signal is encoded in terms of the spatial frequencies of the object using **phase-encoding** and **frequency-encoding gradients**
- every spatial frequency that exists within the image is sampled prior to Fourier transform (known as ‘k-space’)
- inadequate or erroneous k-space sampling leads to image artefacts

Gradients

- Gradient coils are contained within the scanner bore to produce a linear variation of magnetic field strength across the useful imaging volume
- Spatial localisation of the MR signal requires the use of three orthogonal linear magnetic field gradients
- Gradient pulses in conventional pulse sequences are trapezoidal in shape with a sloping rise, followed by a flat plateau and a sloping fall



Gradients

- Strength of a gradient, or how rapidly the magnetic field changes over distance, is expressed in millitesla per metre ($mT\ m^{-1}$)
  - typical values ~45 $mT\ m^{-1}$

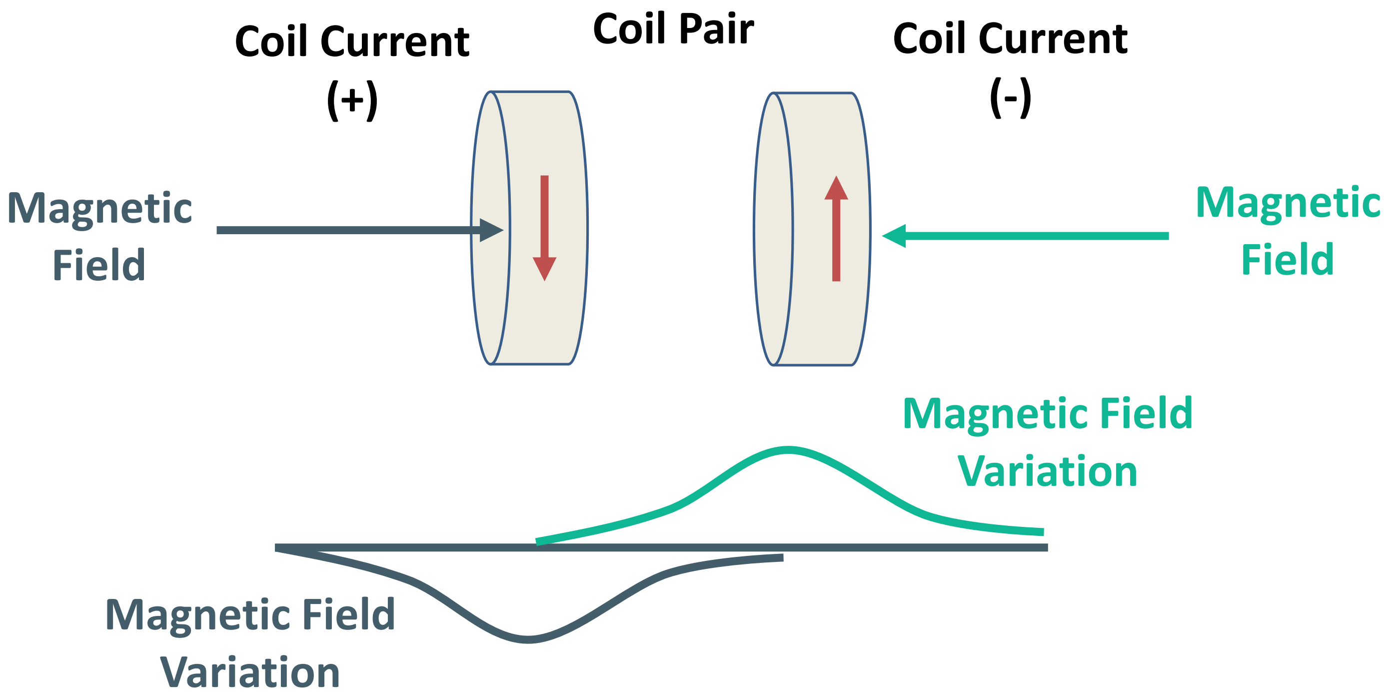
- Gradient rise time, or how rapidly the field changes with time from zero to the peak amplitude, is usually expressed in microseconds (µs), with typical values from 1000 µs down to 200 µs

- The gradient slew rate is calculated by dividing the peak gradient amplitude by the rise-time. Typical slew rates are in the range of 150-200 $T\ m^{-1}\ s^{-1}$

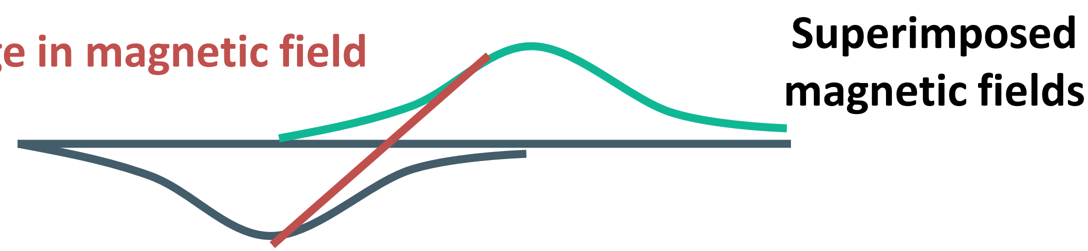
Table 3-23: Gradient information

Component	Specification
Gradient type	Non resonant, actively shielded, rapidly switching
Peak Amplitude	80 mT/m
Slew Rate	200 T/m/s
Rise time to Maximum Amplitude	400 microseconds

Gradients

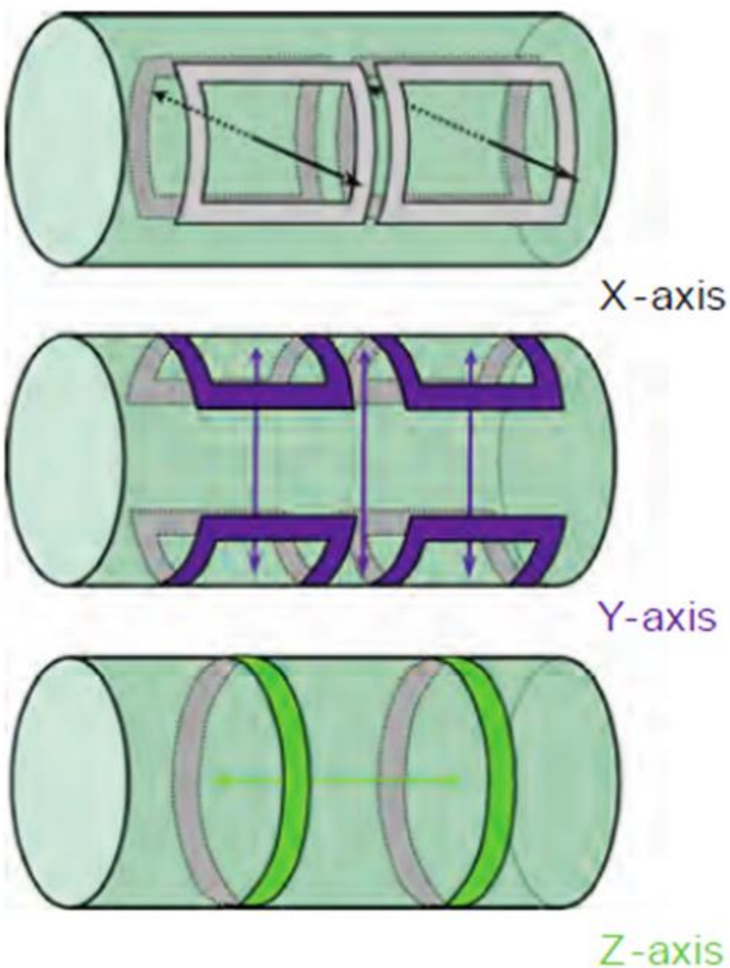


Gradient: Linear change in magnetic field

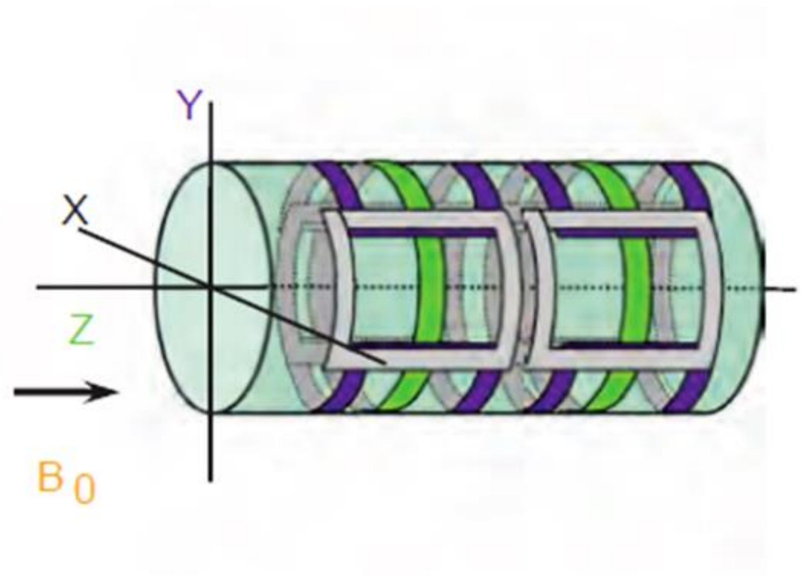


Slice Selection

Individual gradients

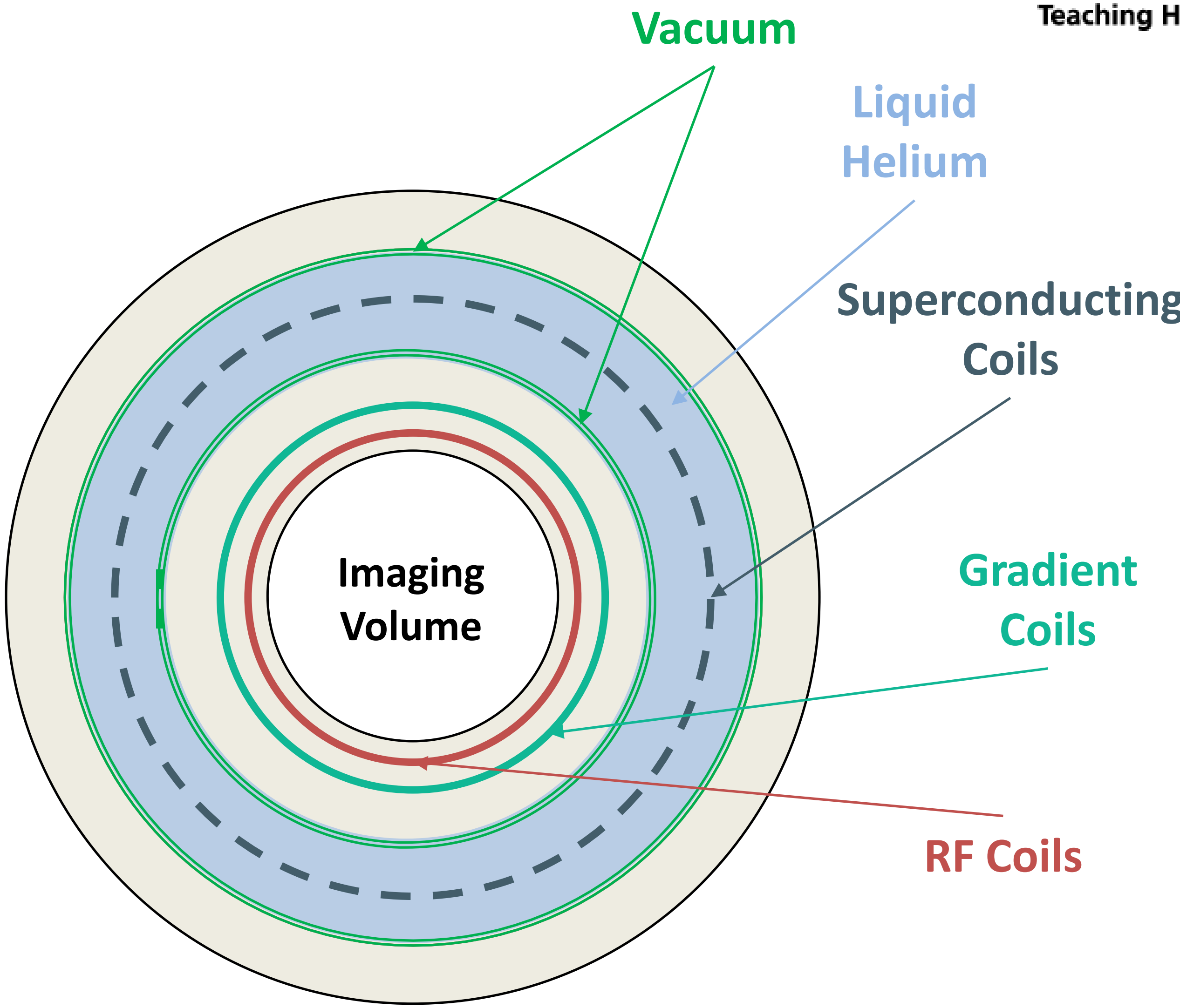


Superimposed gradients

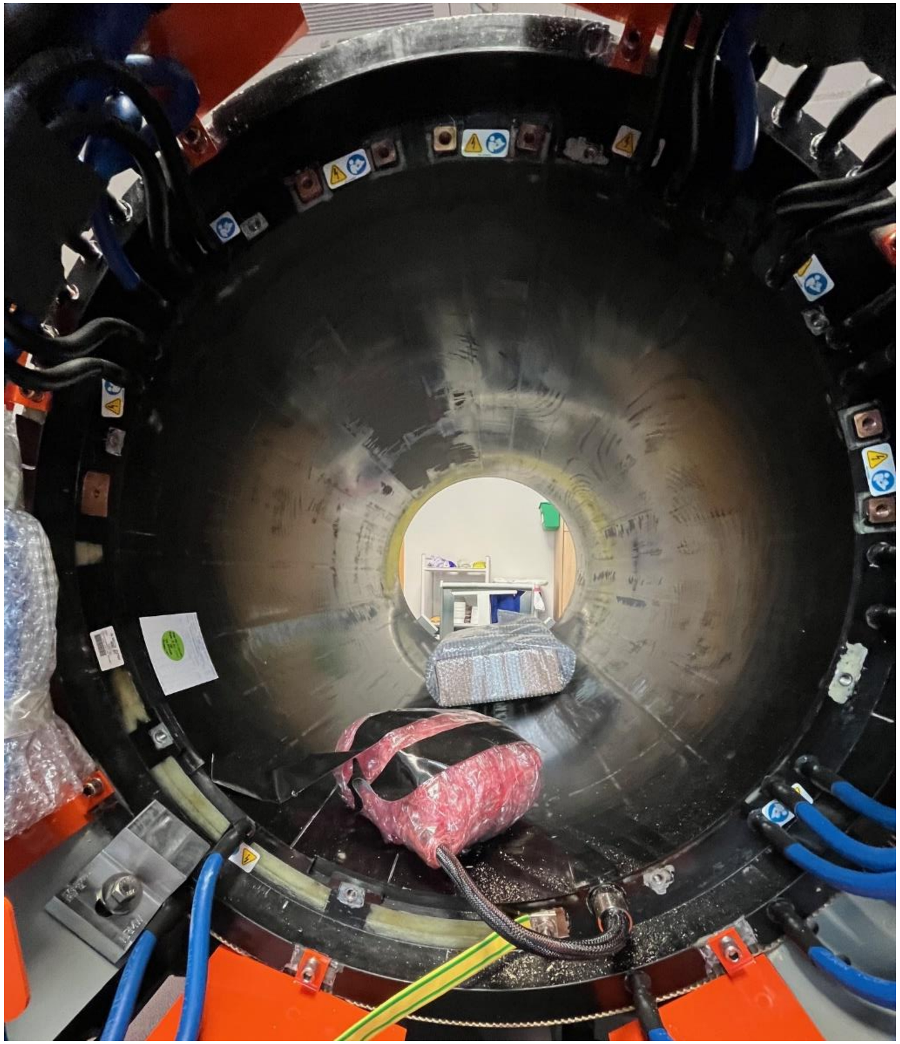


Net gradient = $\sqrt{G_x^2 + G_y^2 + G_z^2}$

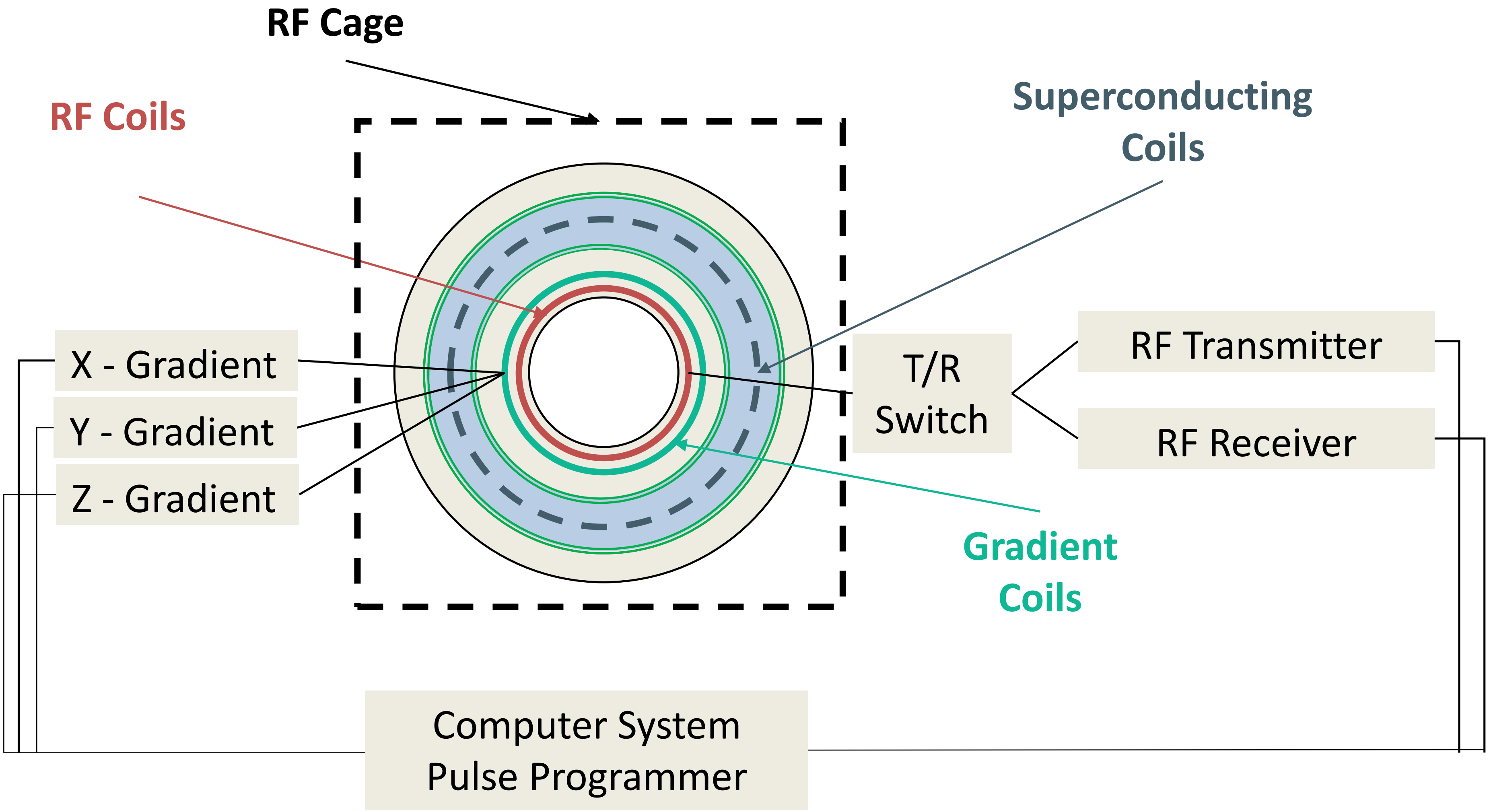
System Design



System Design



System Design



Magnetic Fields in MRI

Component	Amplitude	Pulse Duration
Static magnetic field (B_0)	0.2 – 7T	Always present
Time varying magnetic fields (dB/dt)	0 – 80 mT/m	0 – 10 ms
Radiofrequency magnetic field (B_1)	0 – 50 µT	0 – 10 ms

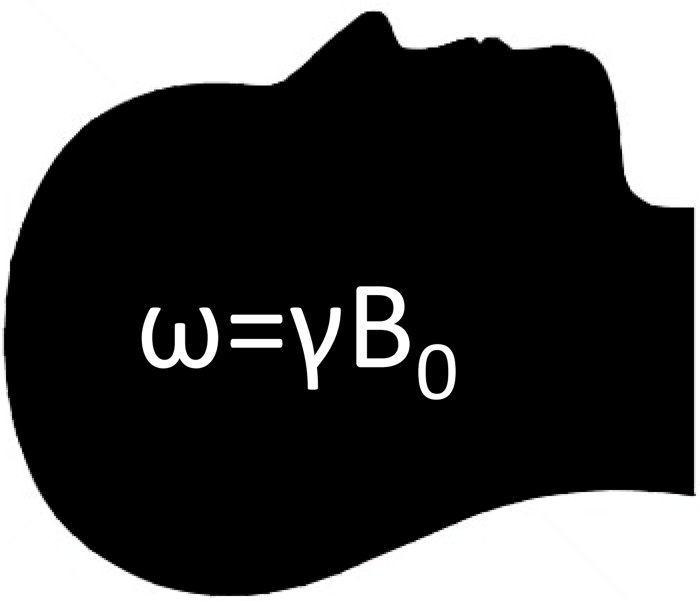
Localisation

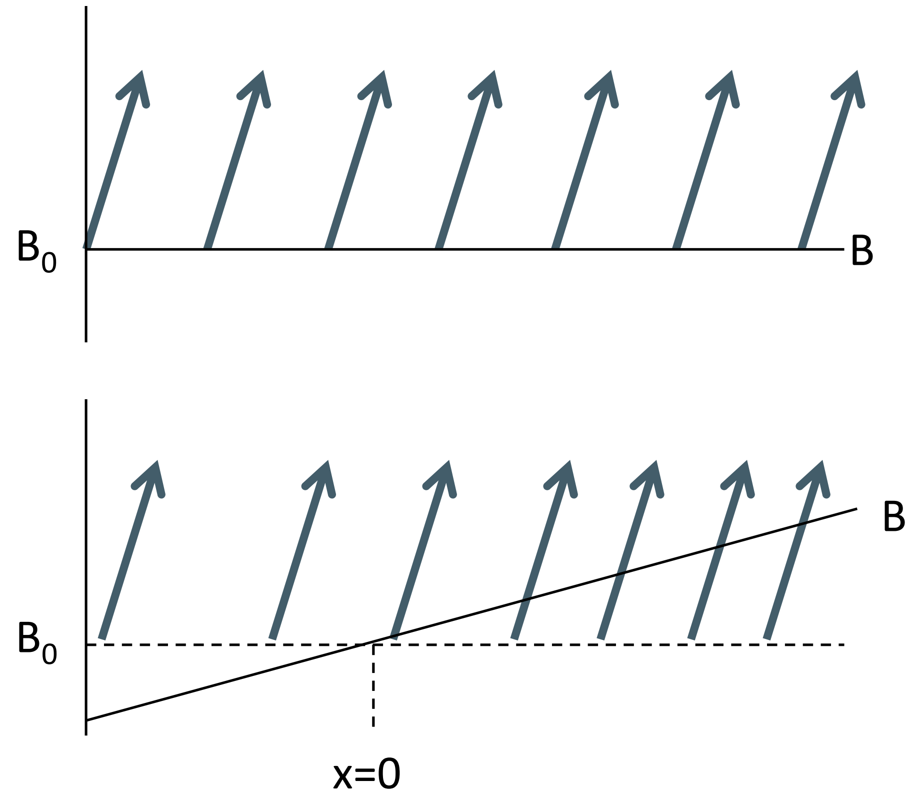
1. Slice selection gradient applied first (Z direction)

Slice Selection

Slice Localisation

- When the term ‘gradient’ is referred to in MRI, it means the addition of spatially linear changes to the B_0 magnetic field

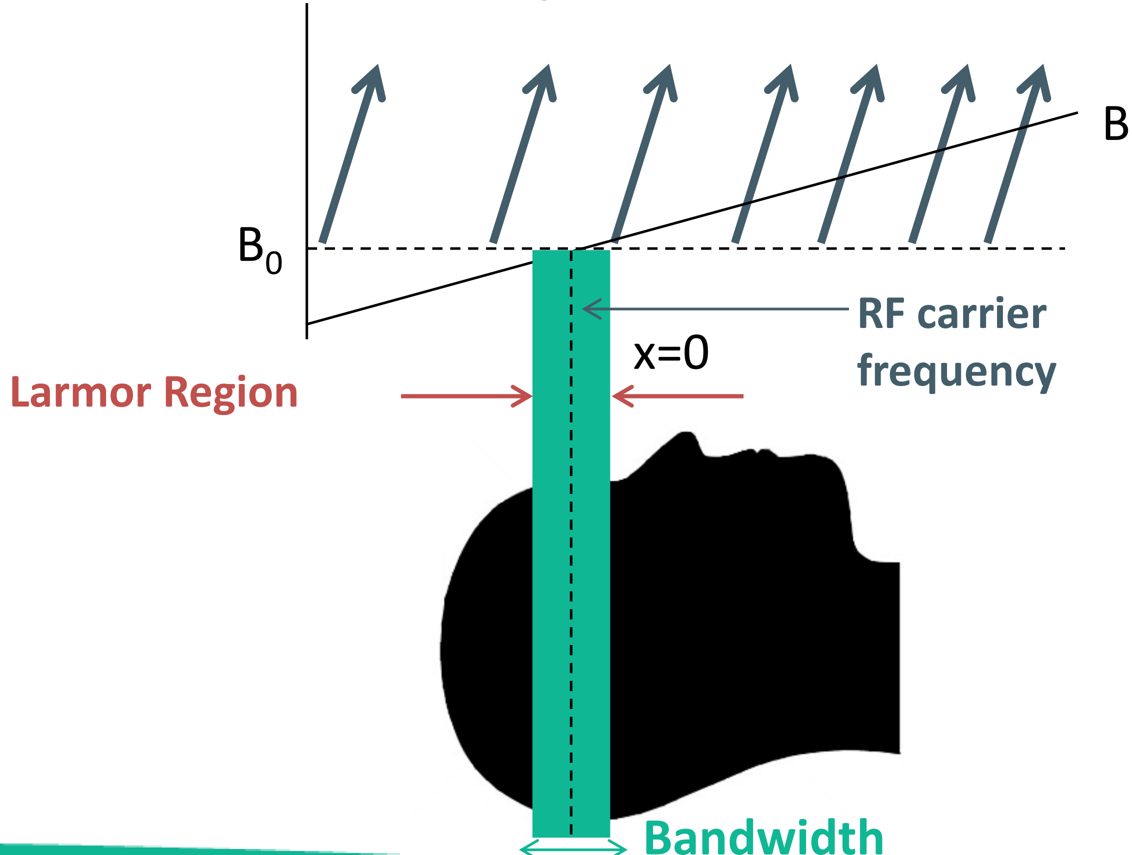




Slice Localisation

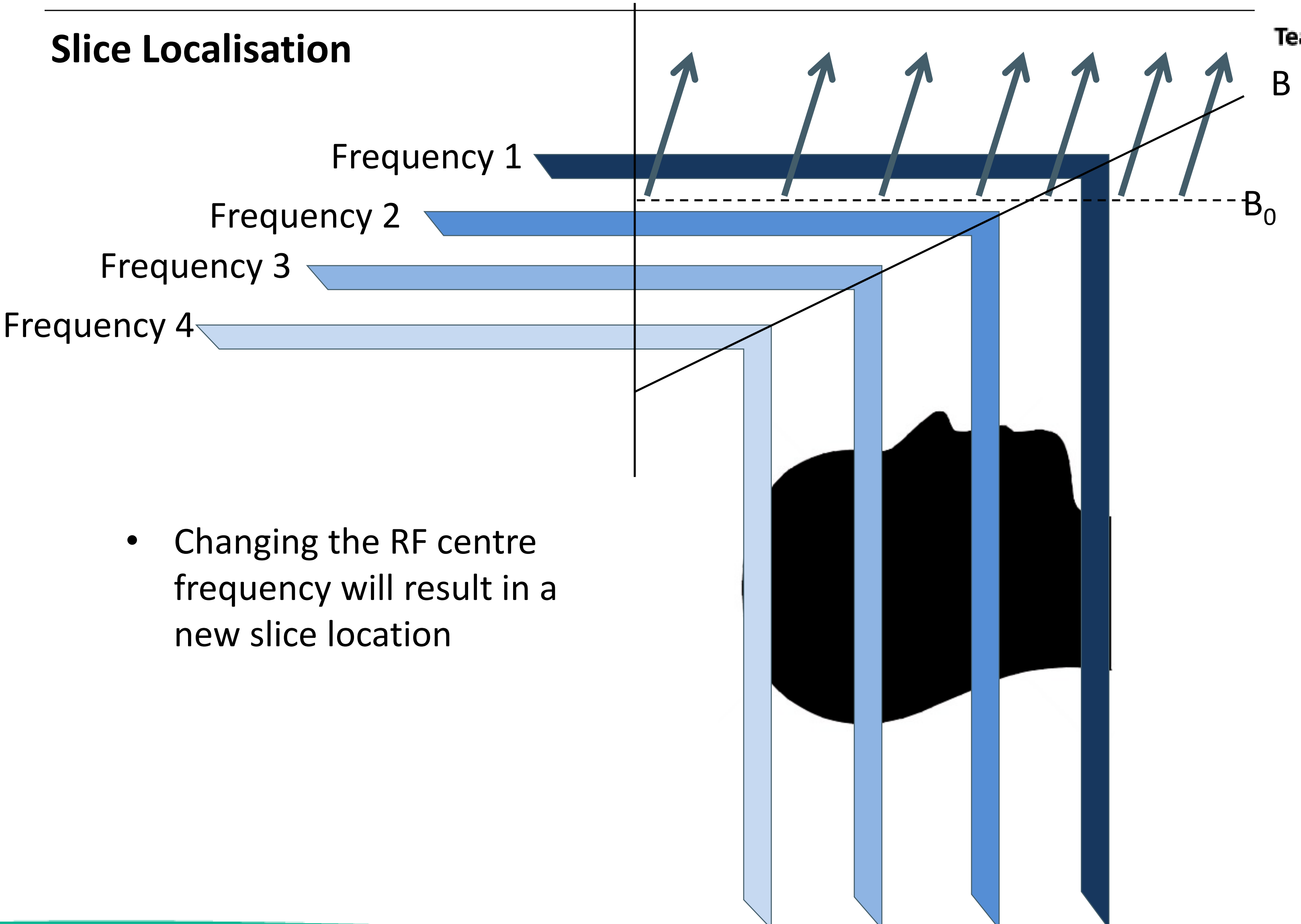
- MR scanners have 3 sets of coils which allow slice selection in any plane

- The magnitude of the change to the B_0 magnetic field is measured in mT m^{-1}
    - e.g. Most HUTH scanners have gradients of 45 mT m^{-1}



Slice Selection

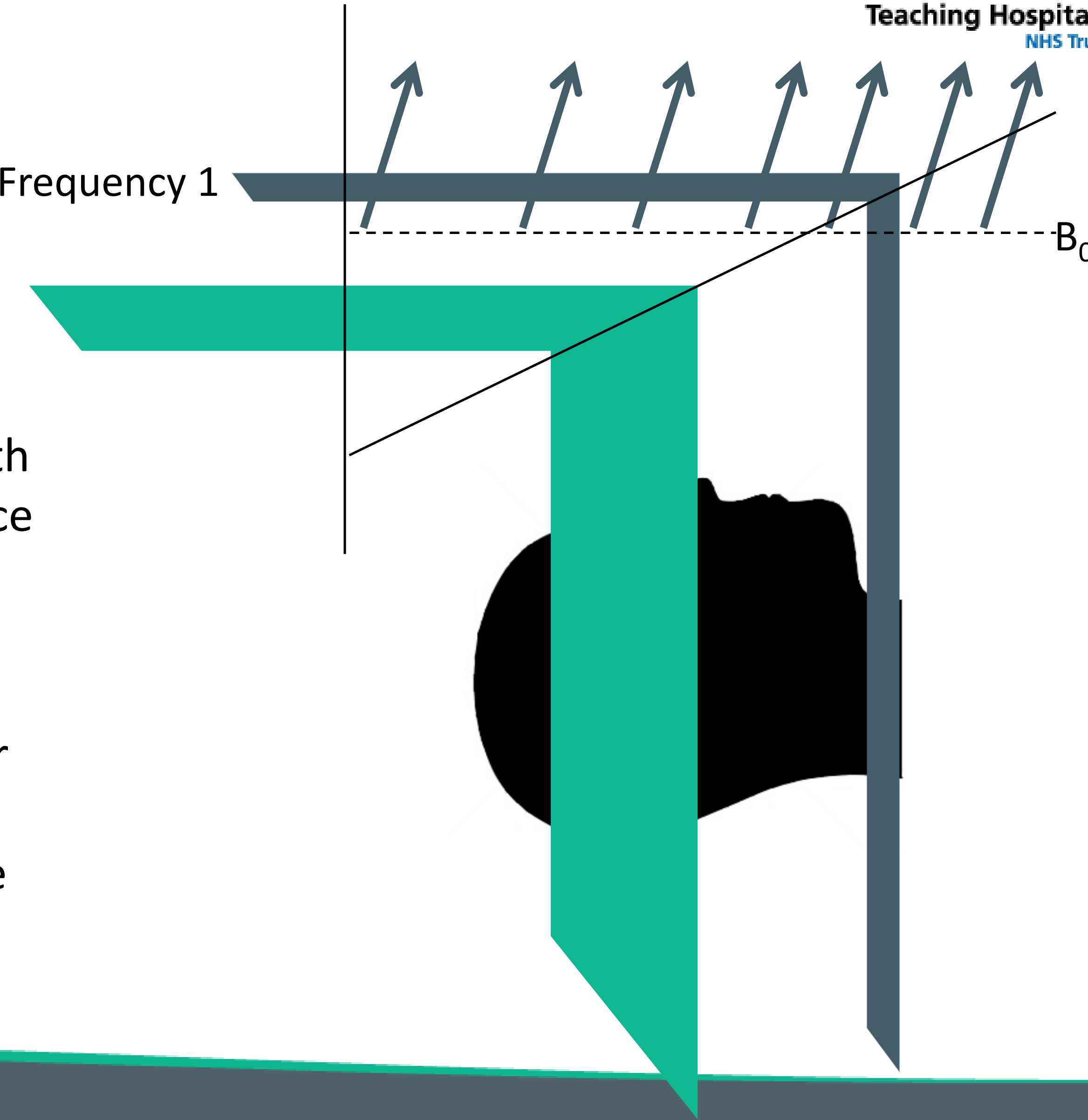
Slice Localisation



- Changing the RF centre frequency will result in a new slice location

Slice Thickness

Slice thickness (z) can be expressed as:
$z = F / \gamma \times G_{ss}$



- A wider carrier bandwidth (F) results in a thicker slice for a fixed gradient

- Alternatively, a shallower gradient (G_{ss}) with fixed carrier bandwidth can be used for thicker slices

γ is the gyromagnetic ratio (For 1H, γ=42.58 MHz/T)

What is k-space?

- k-space is an array of digitised MR echoes in the “frequency domain”
- array of numbers representing all the spatial frequencies of the MR image (same size as the final image)
- The readout MR signal is a mixture of RF waves with different amplitudes, frequencies and phases containing spatial information
- To go from k-space to an image requires a 2D inverse Fourier Transform (2DFT)
- The individual points in k space do not correspond to individual pixels in the image
- Each k-space point contains spatial frequency & phase information about **every** pixel in the final image
- Conversely, each pixel in the image maps to **every** point in k-space

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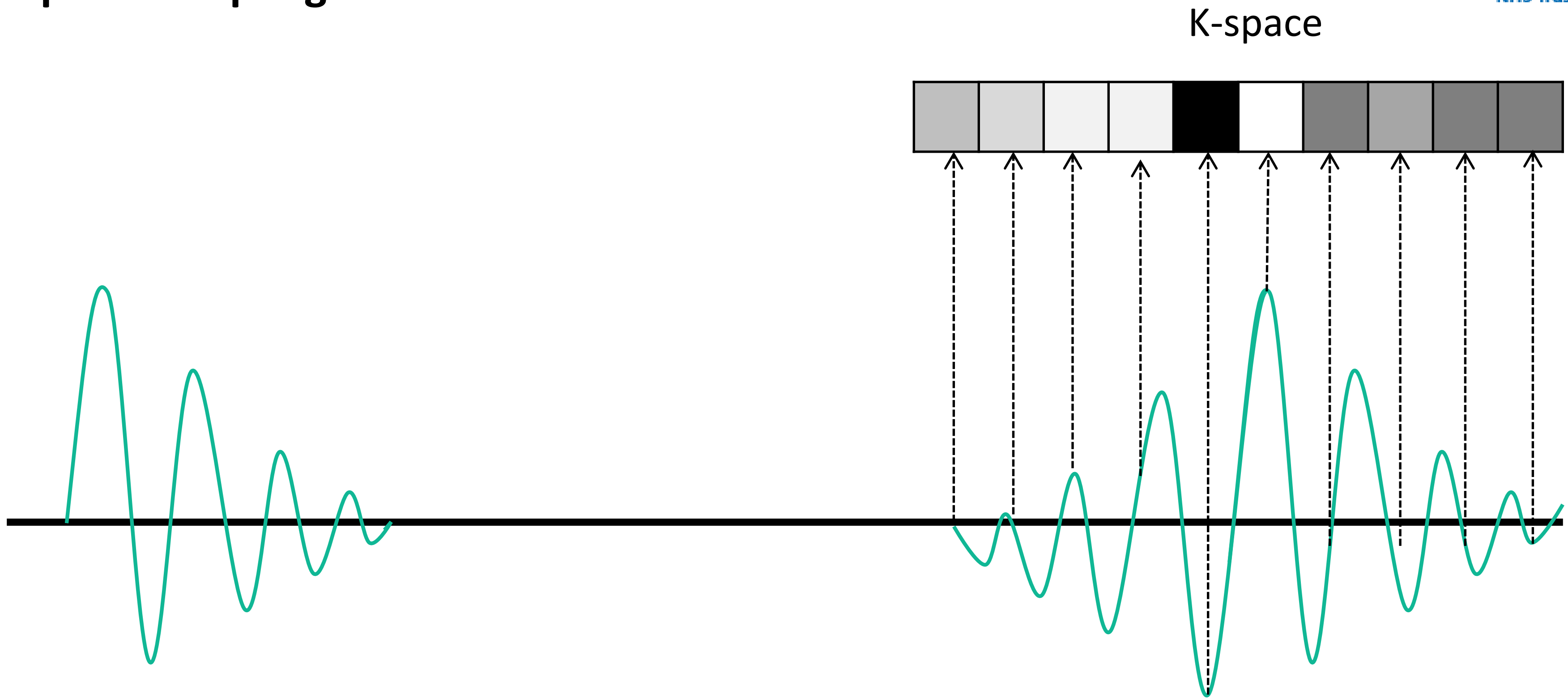
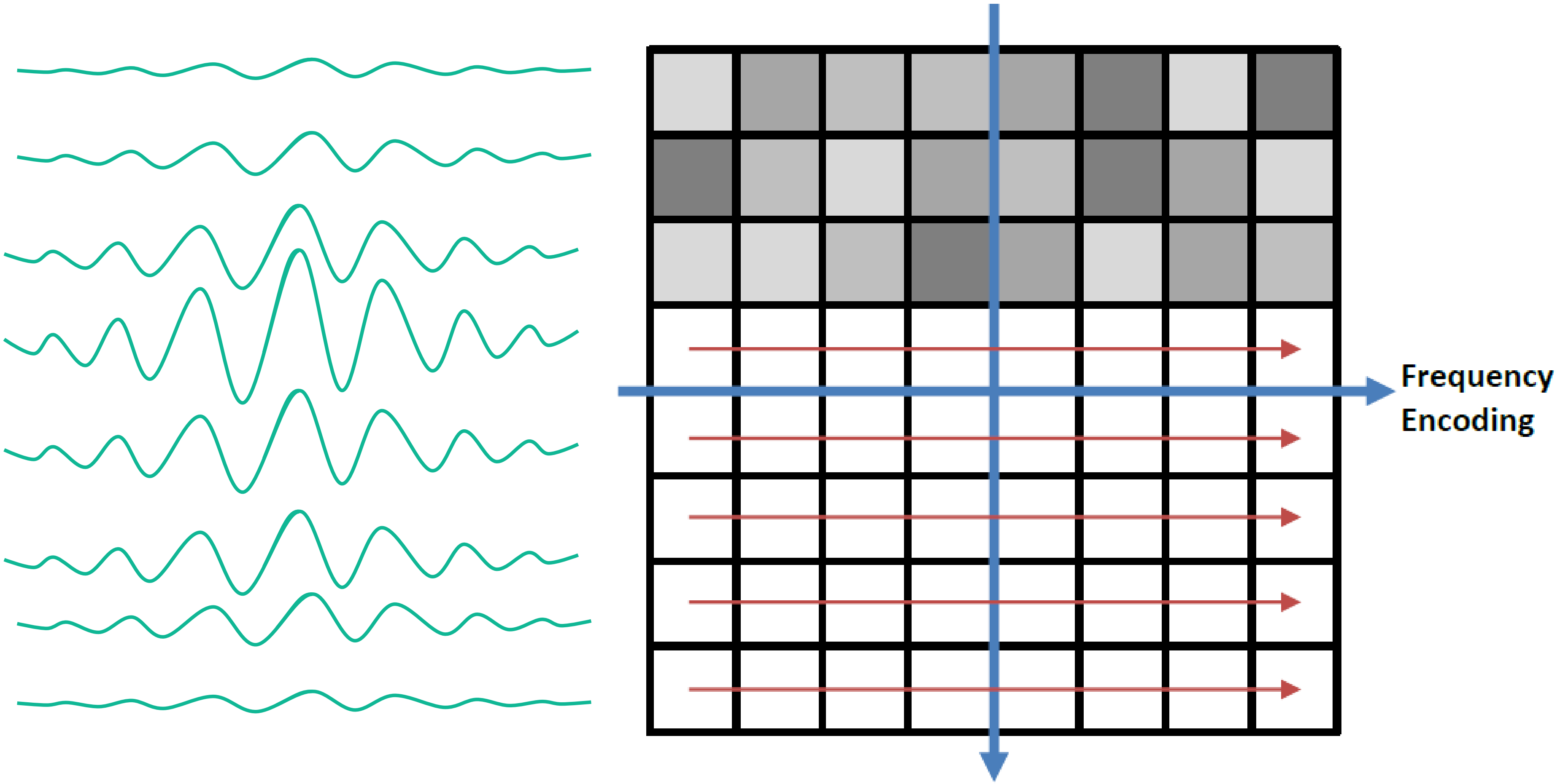
K-space Sampling



- In MRI, the echo rather than the FID is sampled for k-space

K-space Sampling



Localisation

1. Slice selection gradient applied first (Z direction)

2. Phase encoding gradient subsequently applied (Y-direction)

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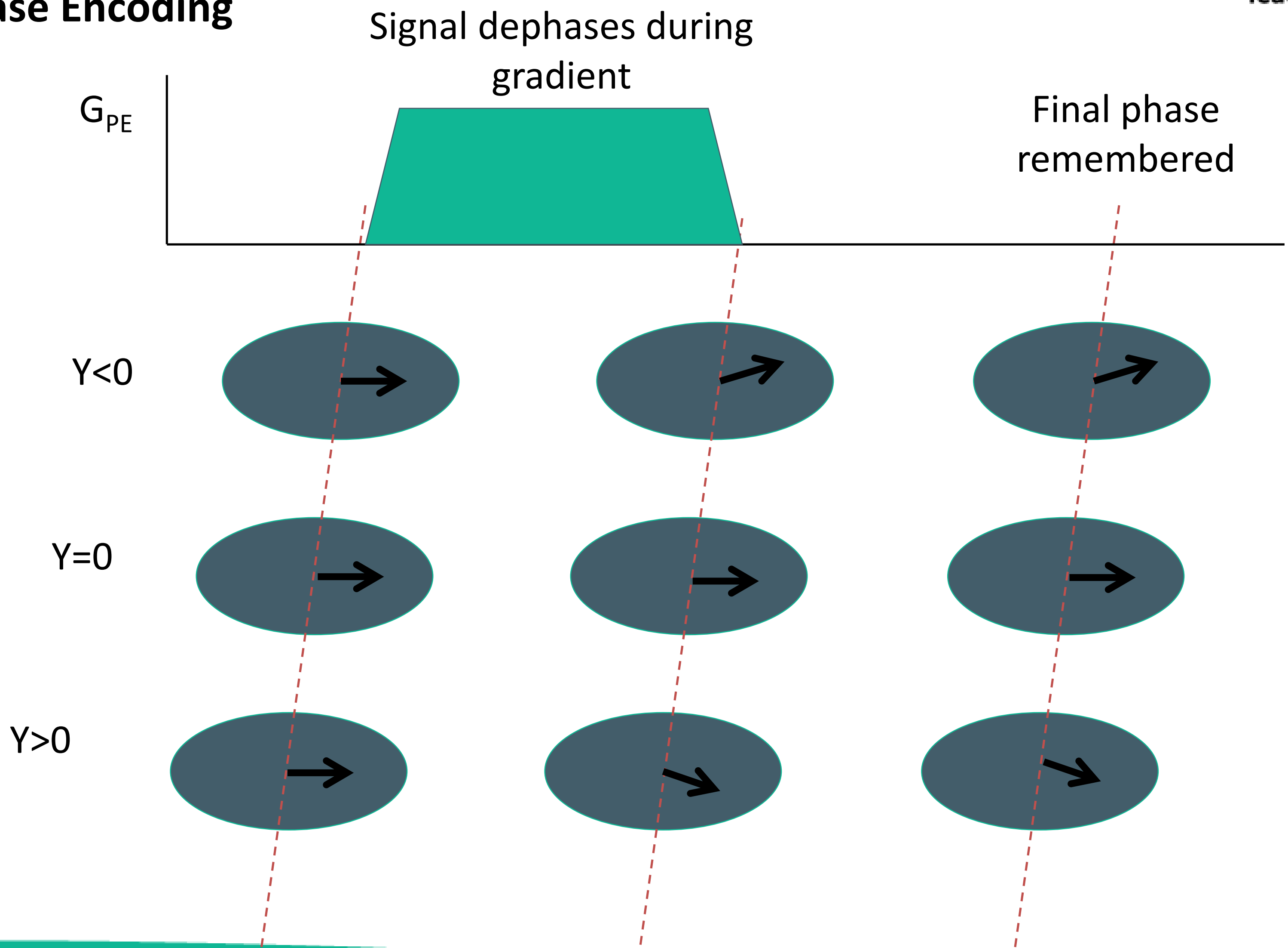
Phase Encoding



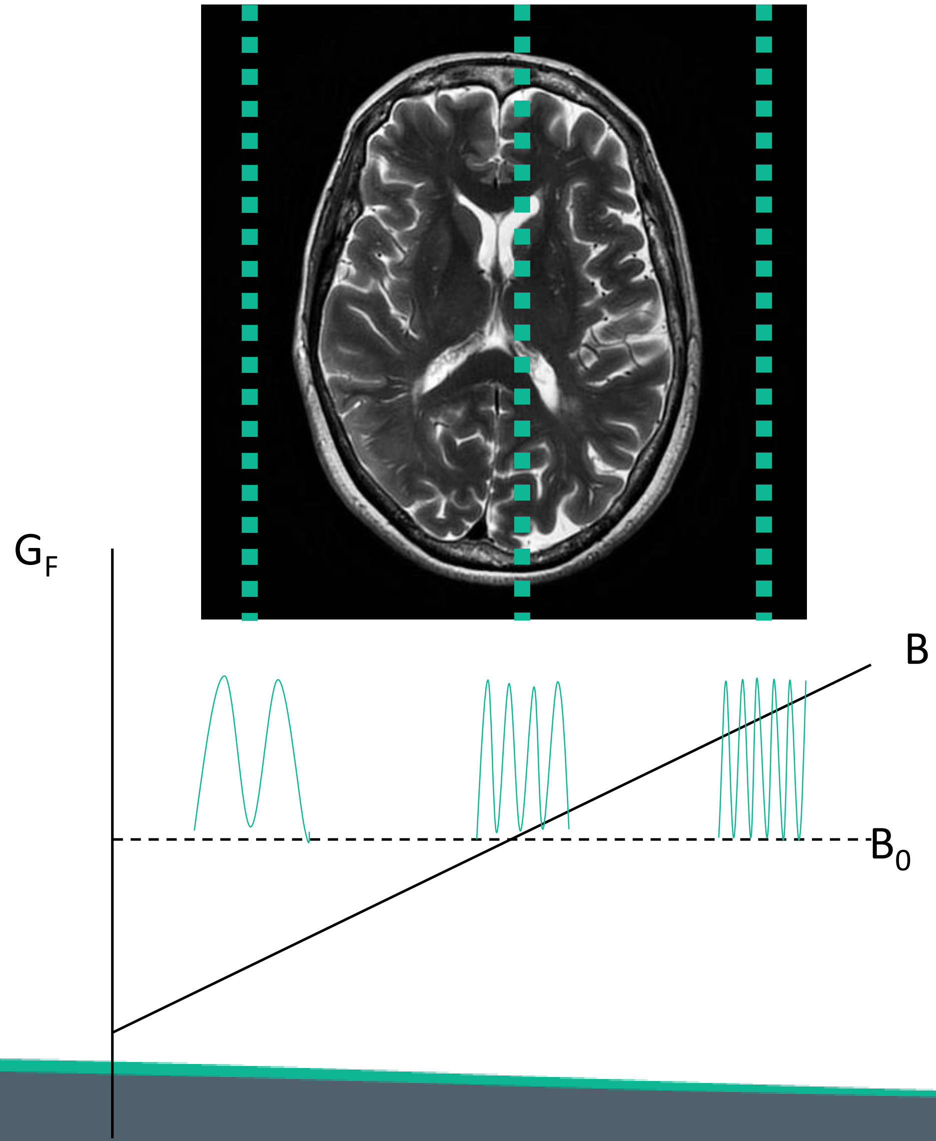
Localisation

1. Slice selection gradient applied first (Z direction)

2. Phase encoding gradient subsequently applied (Y-direction)

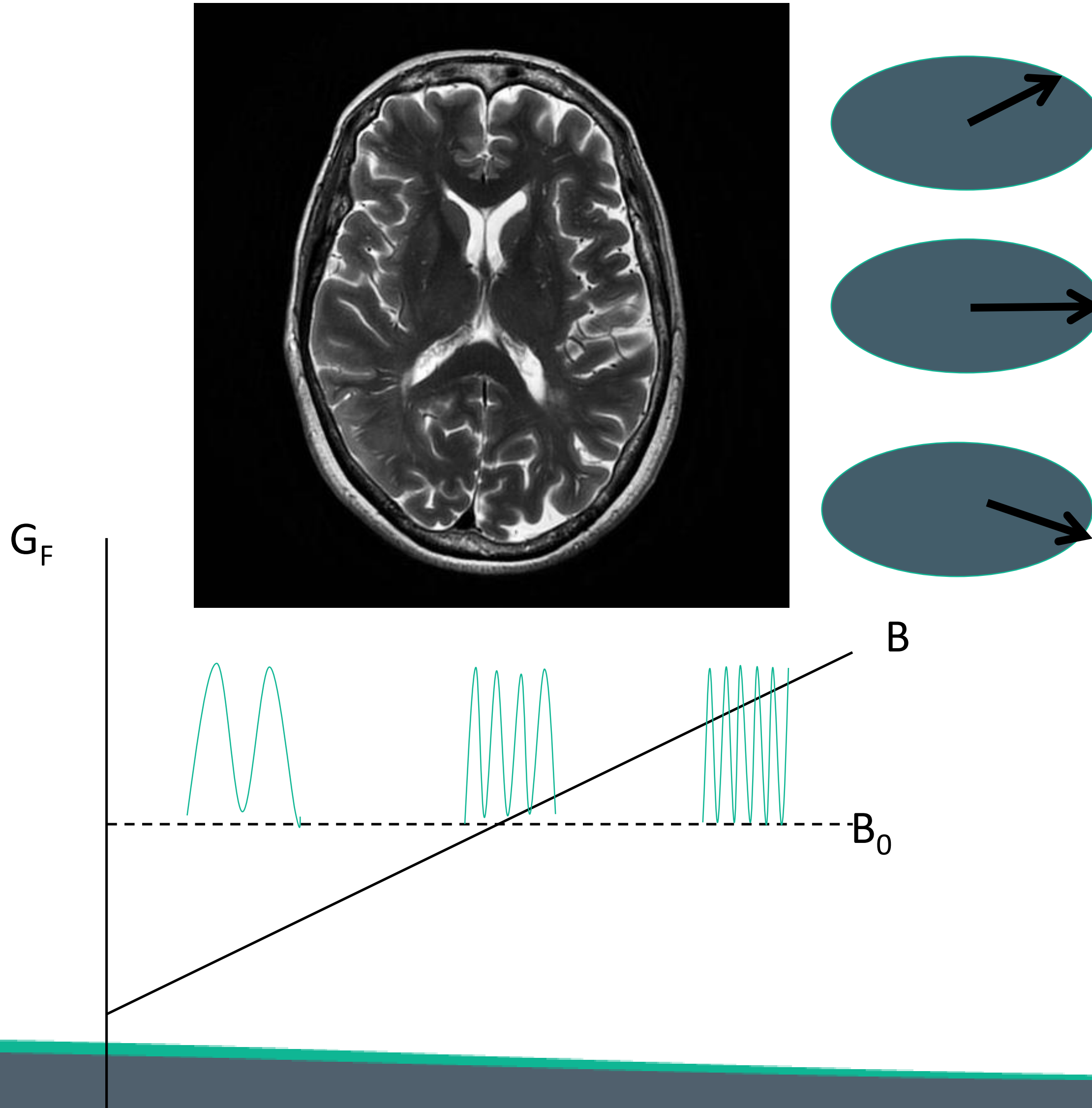
3. Frequency encoding gradient applied last (X-direction)

Frequency Encoding



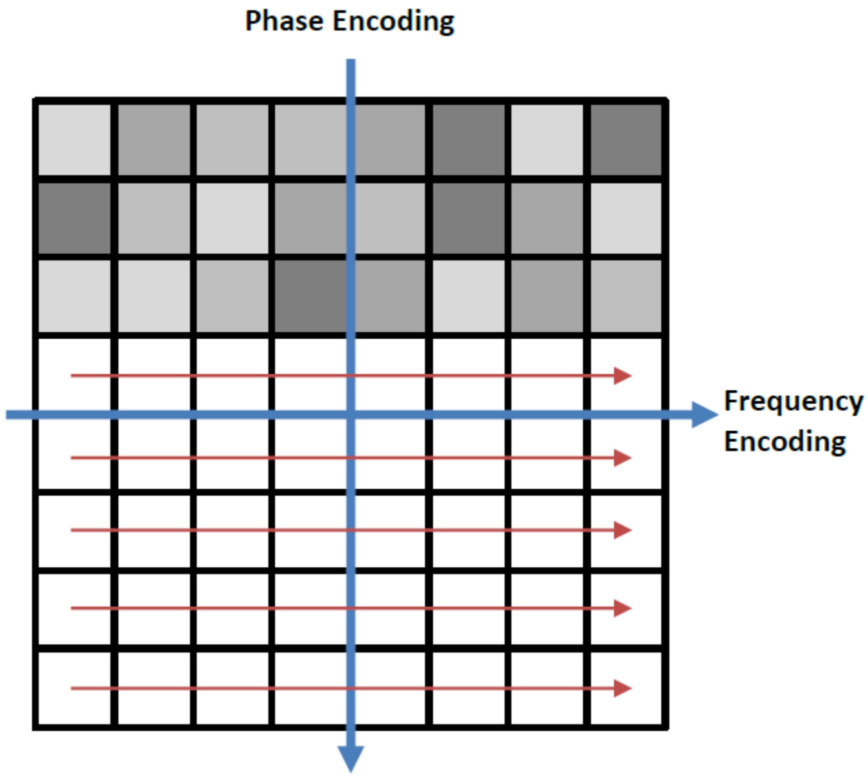
Each pixel has a finite width, so actually contains a small range of frequencies (***pixel bandwidth***) rather than just a single frequency

Frequency & Phase Encoding



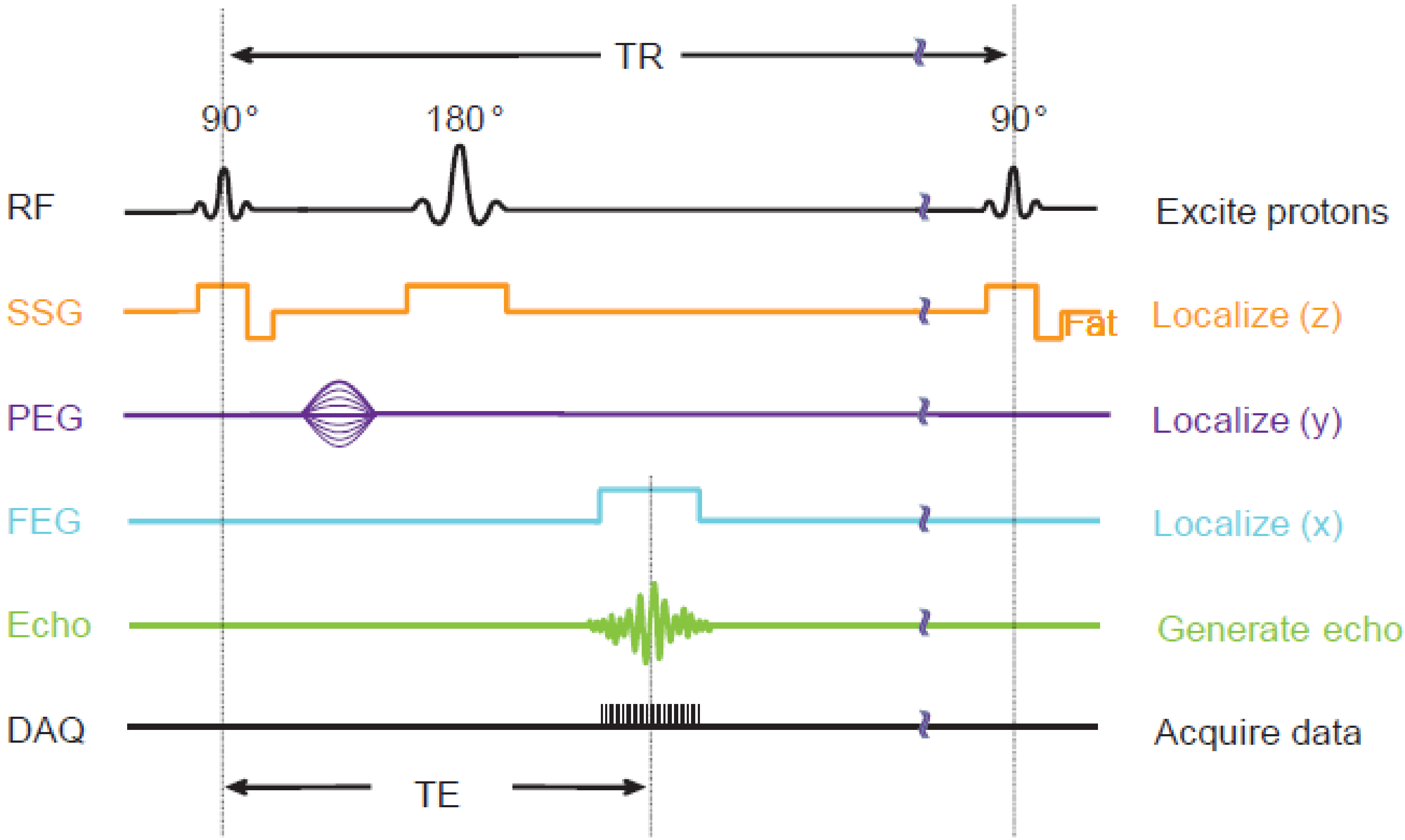
Phase Encoding

- The location that data is stored in k-space depends on the strength and duration of the gradients.
  - If no gradient applied, the location is at the centre of k-space
- The higher the gradient strength or duration, the further from the centre of k-space the data will be located
- Simplest way to fill k-space is line-by-line
- One line of k-space is fully acquired at each excitation (containing low and high horizontal spatial frequency information)
- Phase encoding is slow compared to frequency encoding

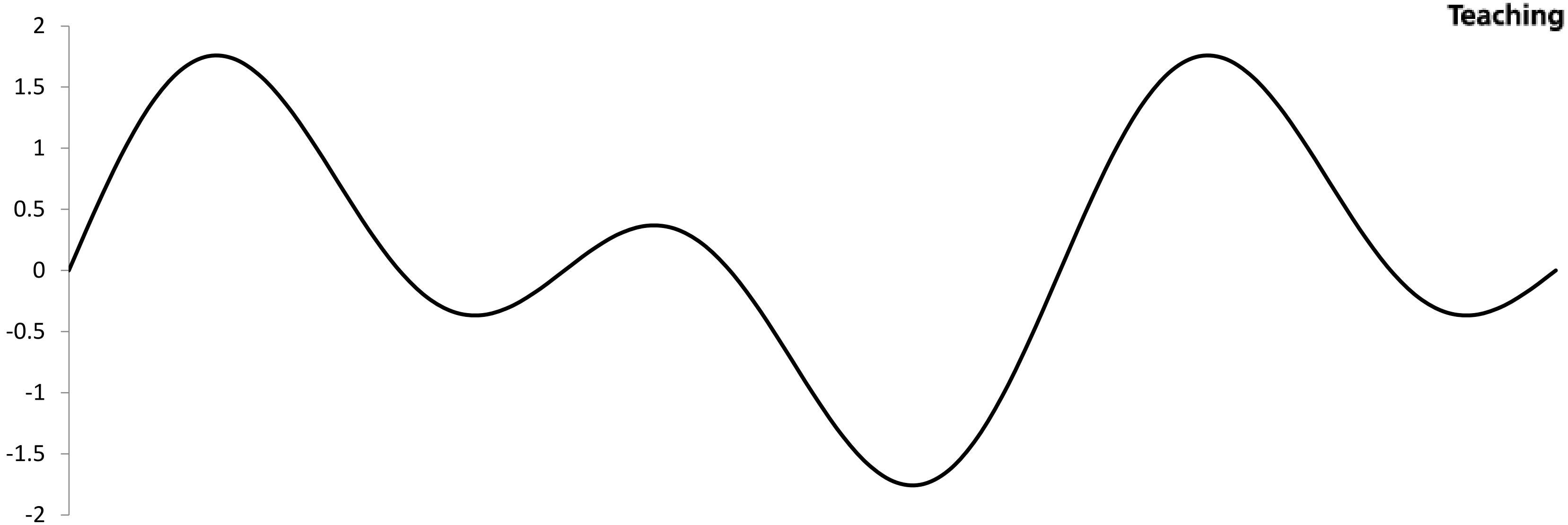


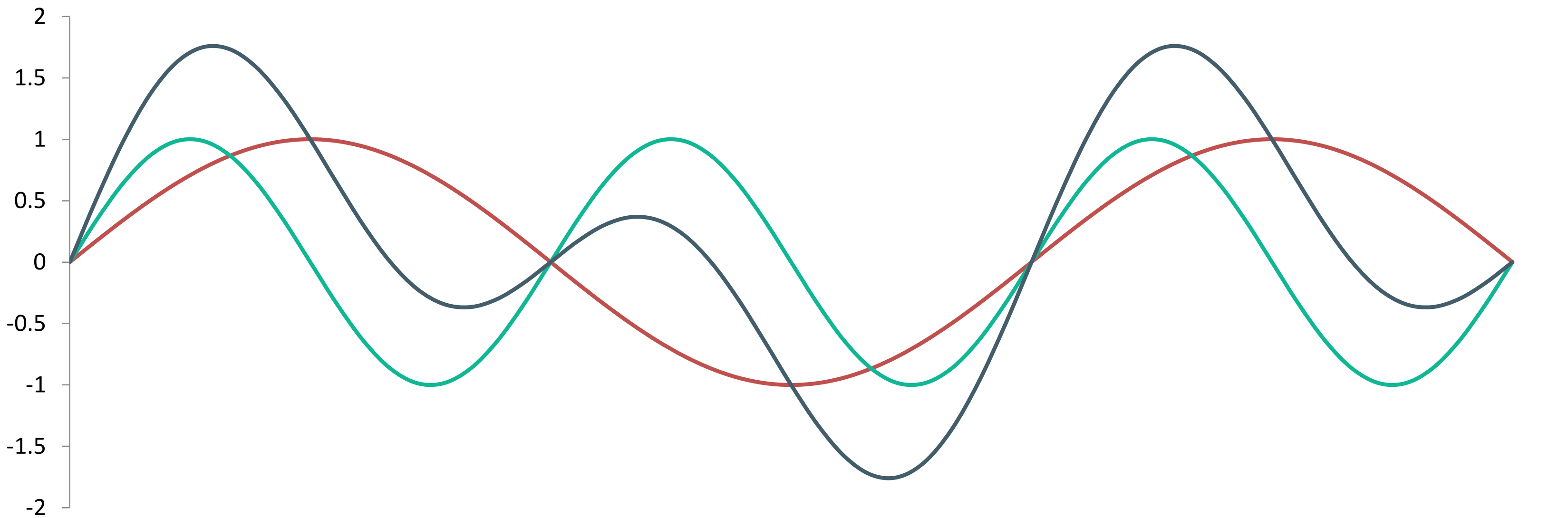
k-space: Relationship between k-space and MR image

Spin Echo Sequence

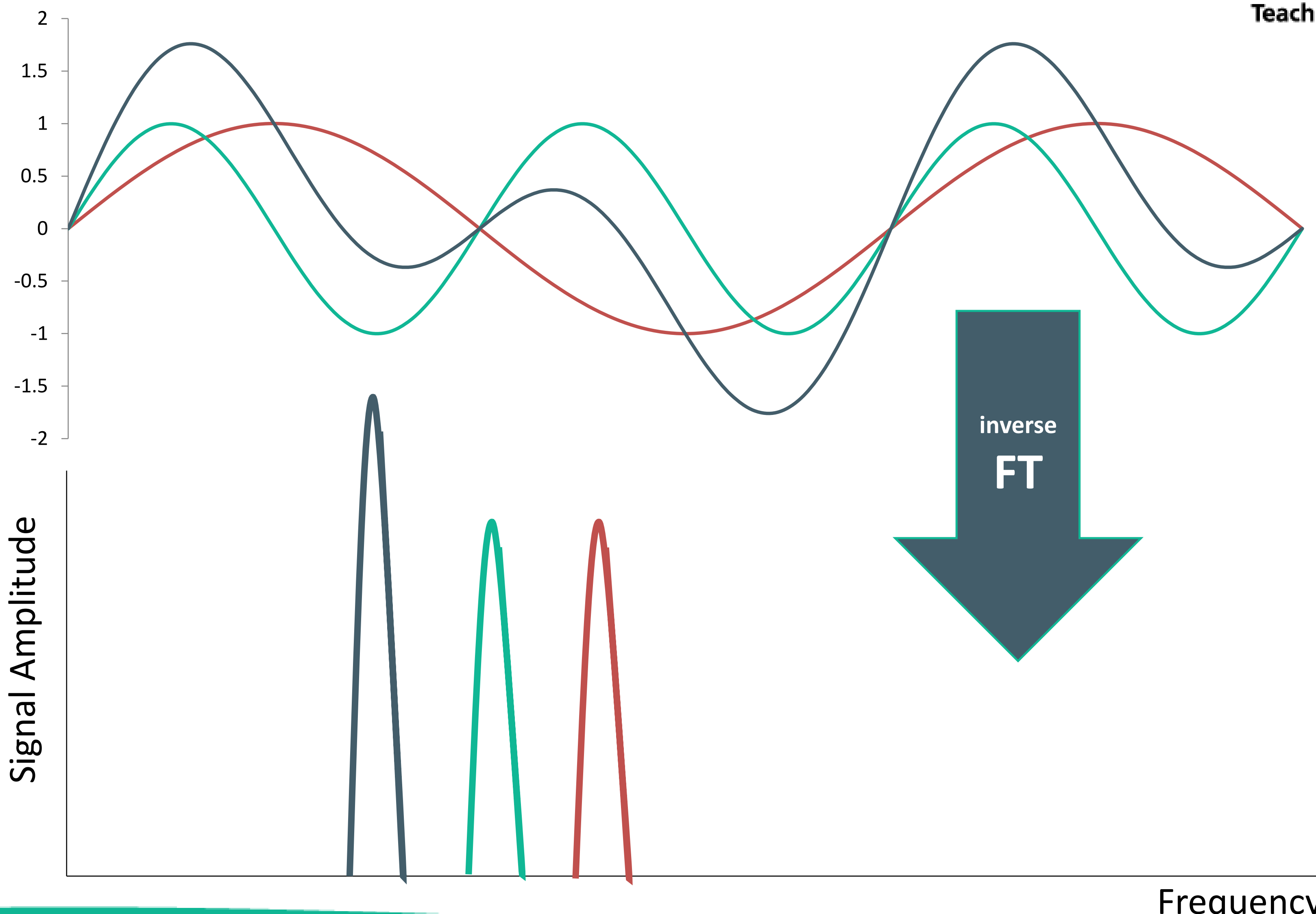


k-space: Relationship between k-space and MR image





k-space: Relationship between k-space and MR image



k-space: Relationship between k-space and MR image

- Spatial frequency is related to the periodicity with which the image intensity values change.

- Image features that change in intensity over **short** image distances correspond to **high** spatial frequencies.

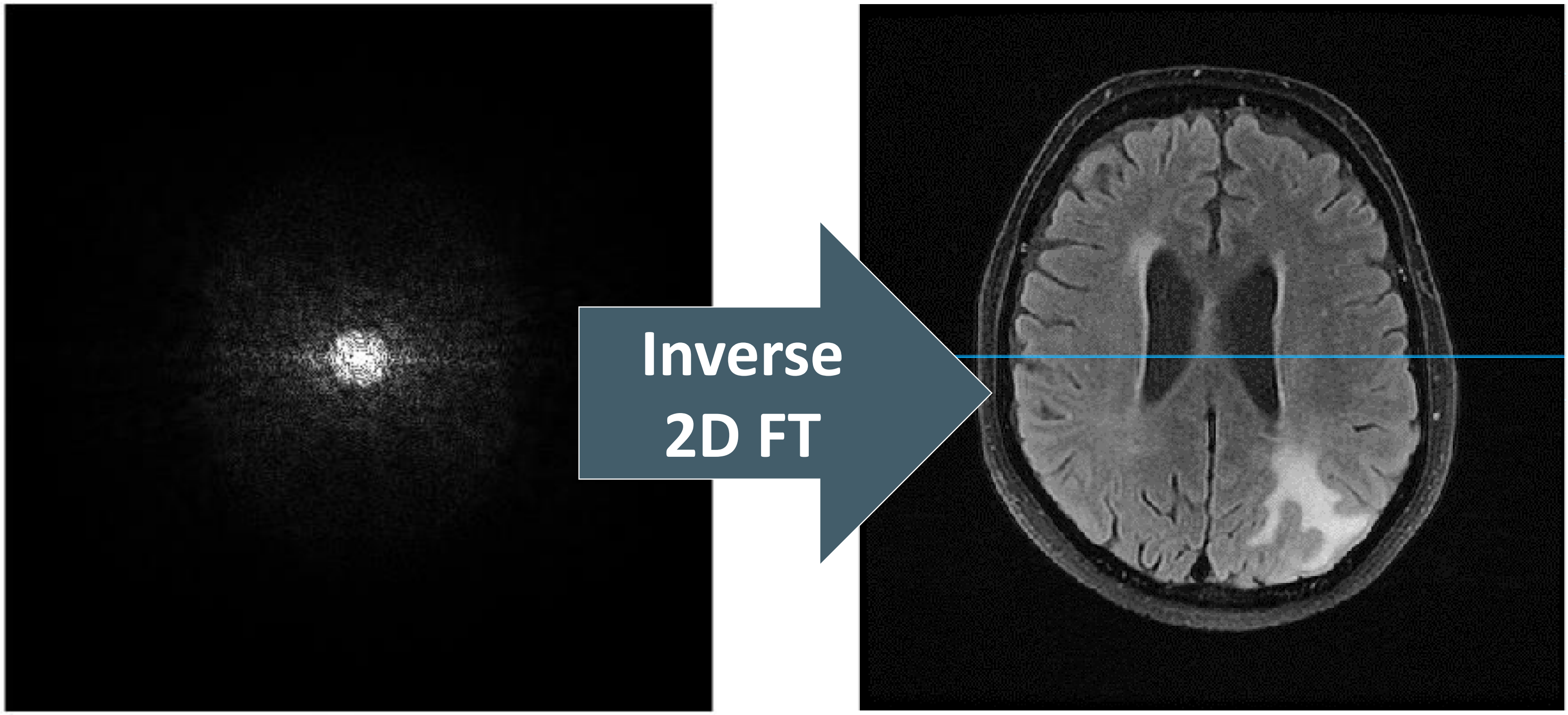
- Image features that change in intensity over **long** image distances correspond to **low** spatial frequencies.

- To decompose a 2D image we need to perform a 2D Fourier Transform

- The echo signal is recorded in quadrature, so each k-space point contains real and imaginary components

- In MRI we look at the Magnitude image = $\sqrt{(Real^2+Imaginary^2)}$

2D Fourier Transforms



Frequency Domain (k-space)

Spatial Domain (image)

k-space: Relationship between k-space and MR image

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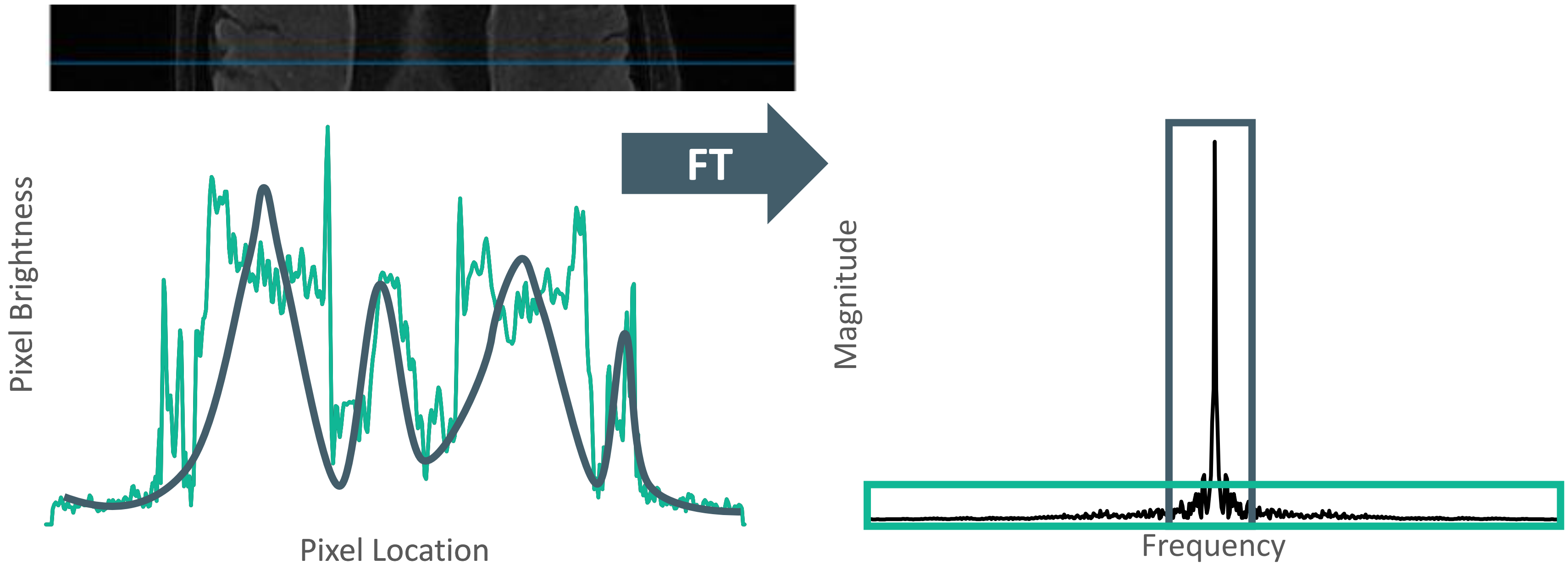


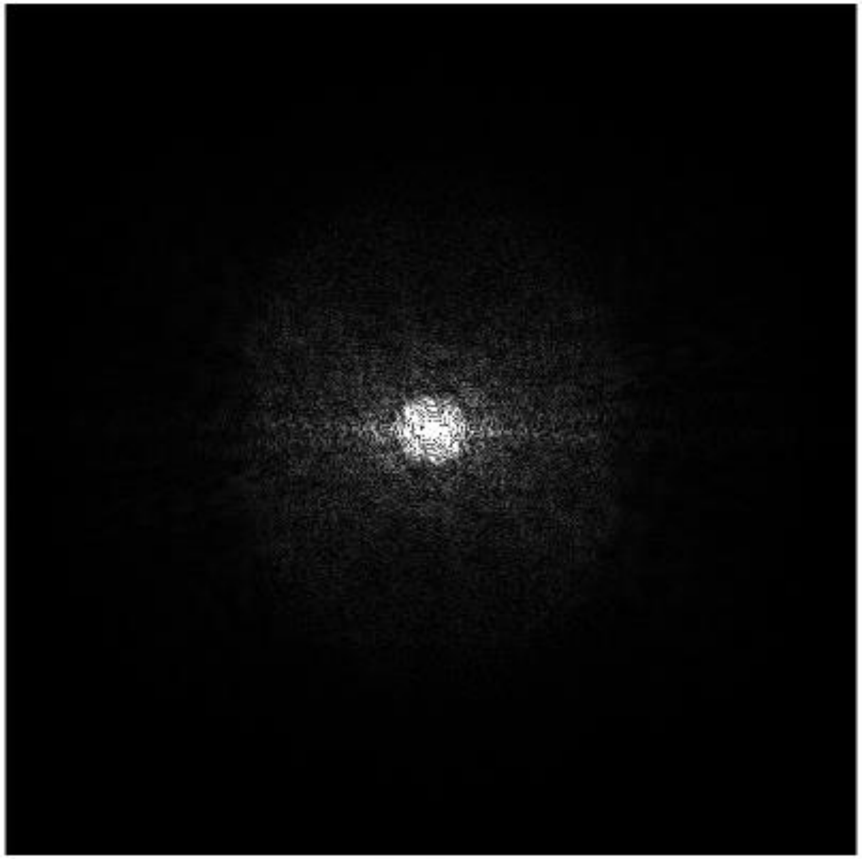
Space Domain

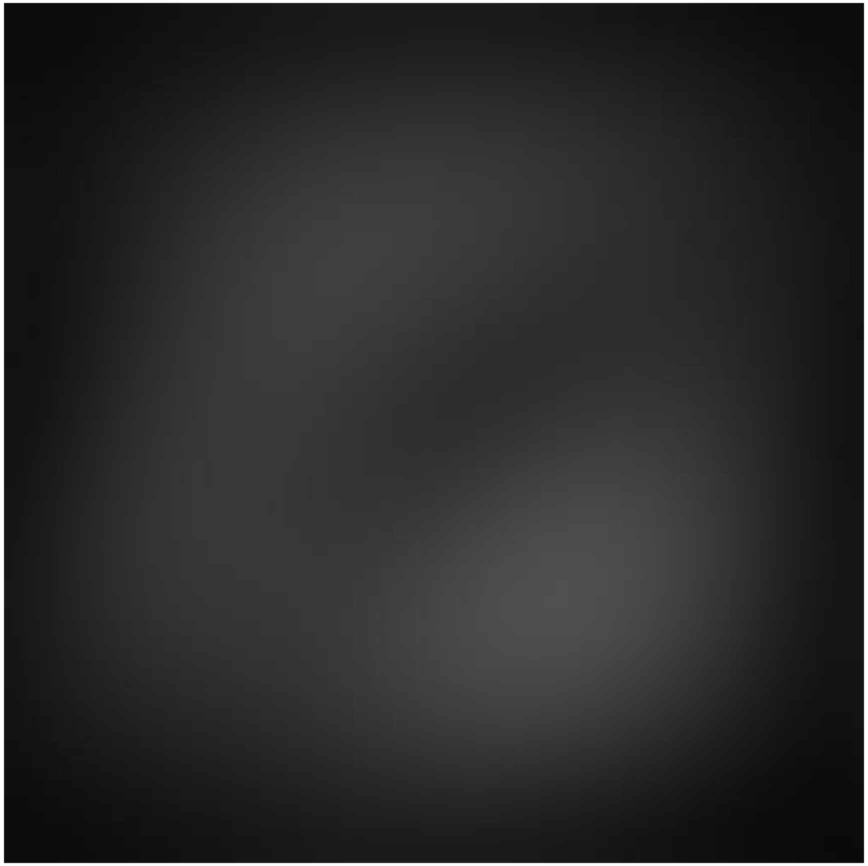
Frequency Domain

- Low spatial frequencies (change in intensity over long distances) are prevailing
- Low spatial frequencies have greatest change in intensity
- High spatial frequencies (change in intensity over short distances) have lower amplitudes
- General shape of the image is described by low spatial frequencies

https://www.imaios.com/en/e-Courses/e-MRI/The-Physics-behind-it-all/2D-Fourier-transform

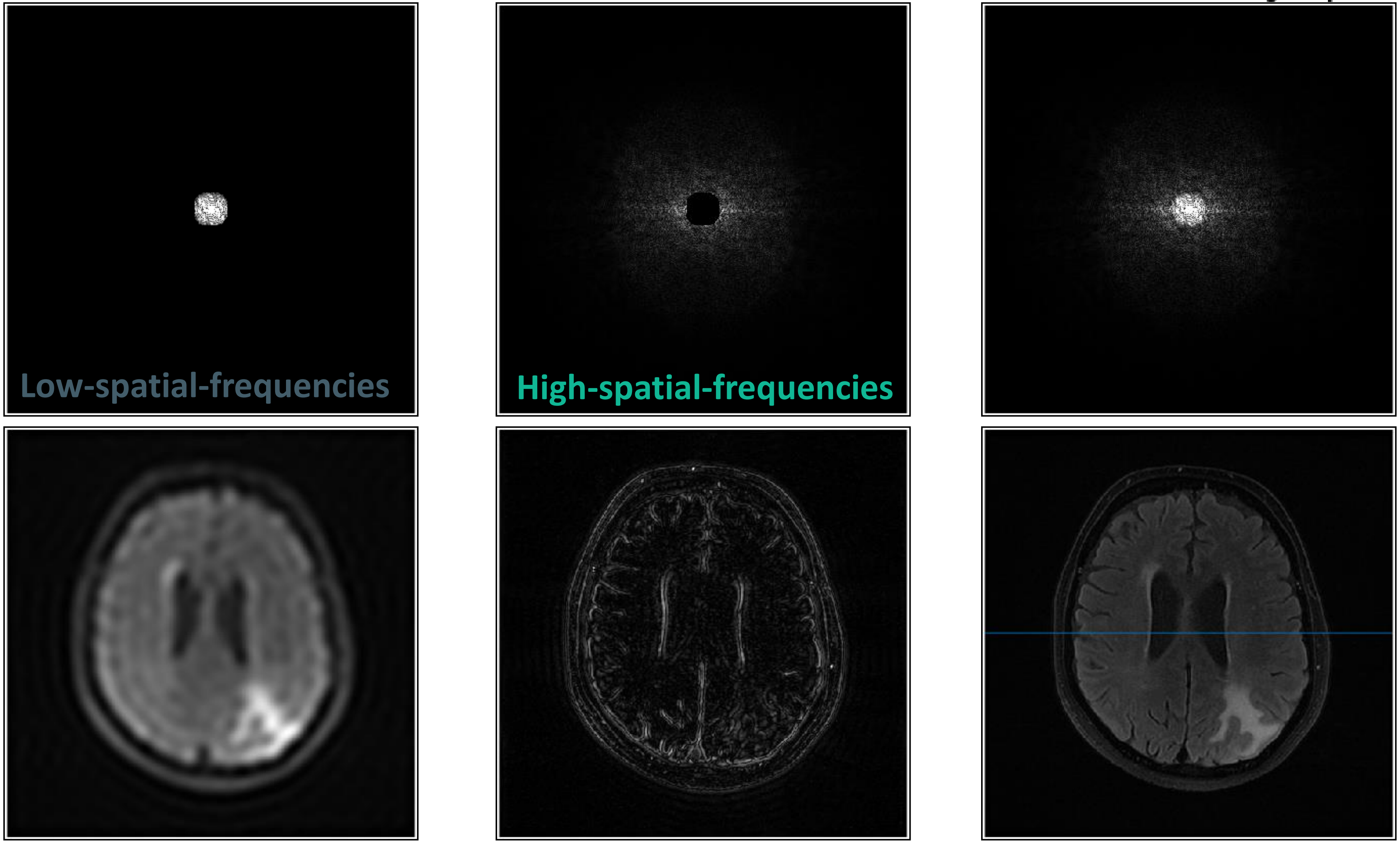
2D Fourier Transforms





Concentric circles of k-space from the centre

k-space: Relationship between k-space and MR image



k-space: Relationship between k-space and MR image

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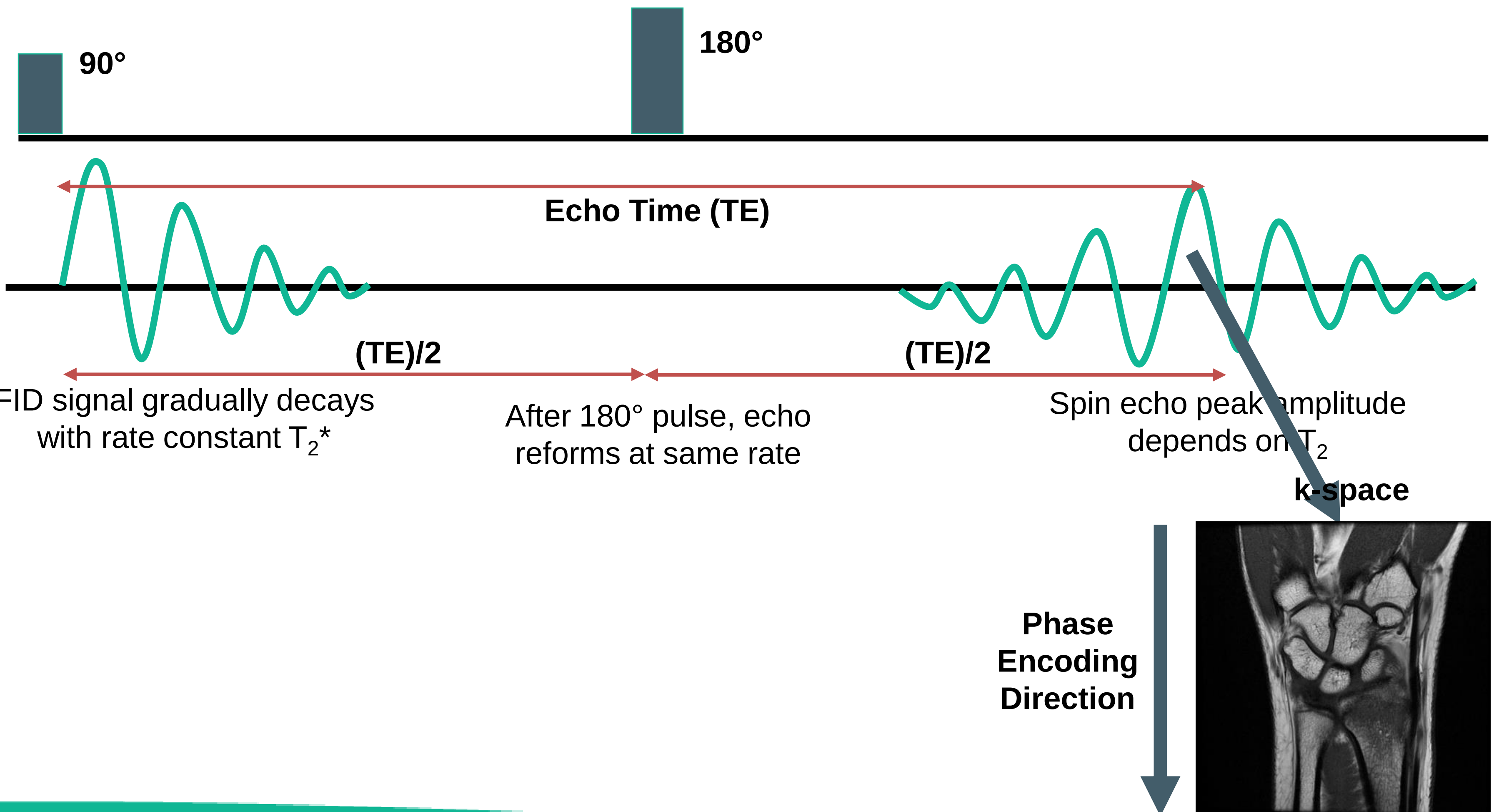
- Spin Echo (SE)
- Simplest pulse sequence



Spin Echo Image acquisition

- **Narrow band RF excitation** pulse simultaneously applied with **the slice select gradient** causing a specific slab of tissue to be excited

- **Transverse magnetisation (M_{xy}) is produced** with amplitude dependence on the saturation of the protons and the angle of excitation

- **Phase encoding gradient is applied briefly**, introducing a phase difference among the protons along the phase encode direction

Spin Echo

- A **refocusing 180-degree RF pulse** is delivered at TE/2 to invert and re-establish the phase coherence of the transverse magnetisation at time TE

- During the echo formation, the **frequency encoding gradient is applied**, generating spatially dependent changes in the precessional frequencies of the protons

k-space: Relationship between k-space and MR image

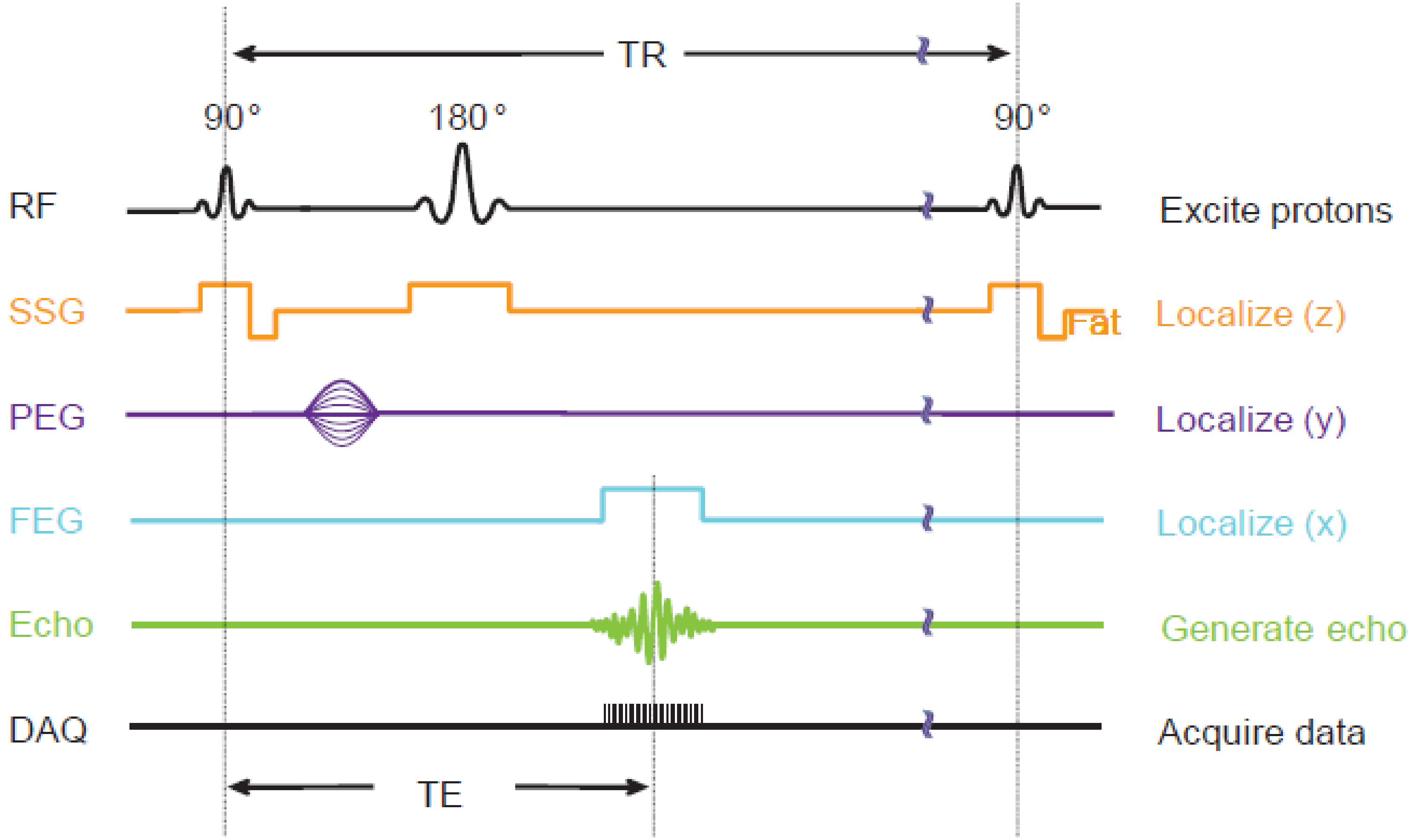
- **Data sampling and acquisition** of the signal occurs simultaneous to the **frequency encoding gradient**

- Data is deposited in the k-space matrix at a row location determined by the strength of the phase encoding gradient

- For each TR, an incremental change of the phase encoding gradient strength sequentially fills each row

- Following the **complete filling of k-space, an inverse Fourier transform** decodes the frequency domain variations in phase for each of the columns of k-space to produce the spatial domain representation - an image!

Spin Echo Sequence



Awareness of different k-space trajectories and their advantages/disadvantages

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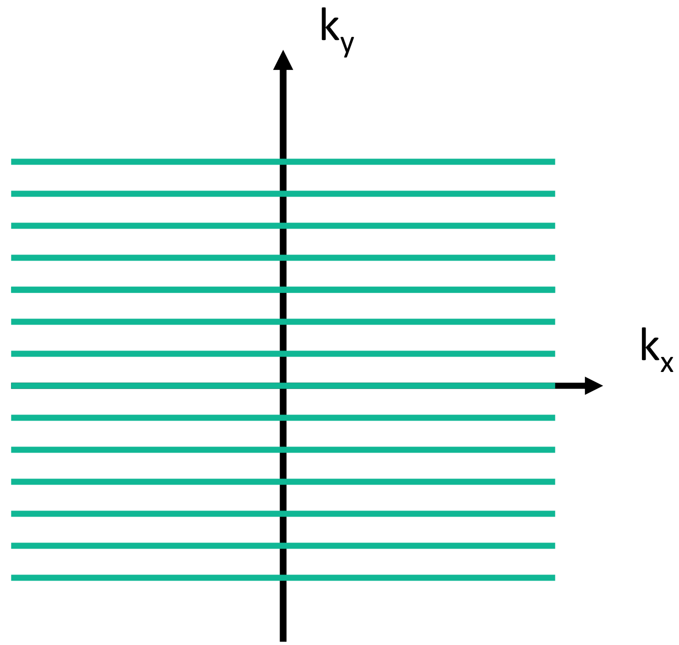
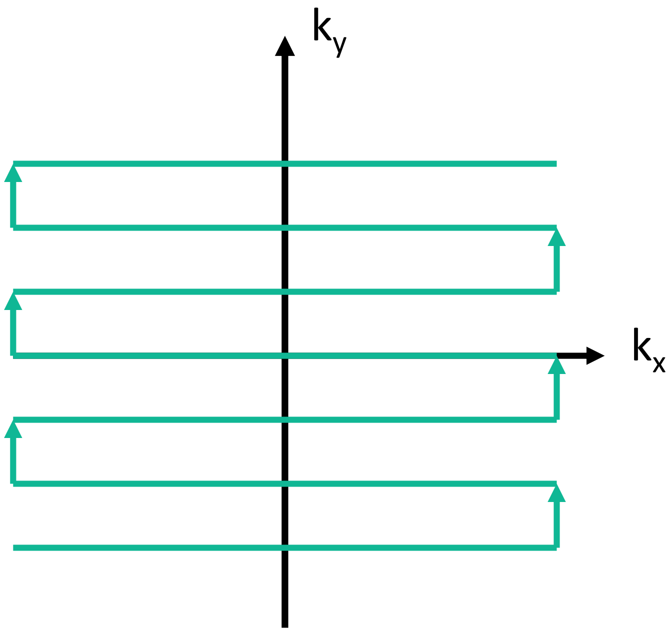
k-space trajectories



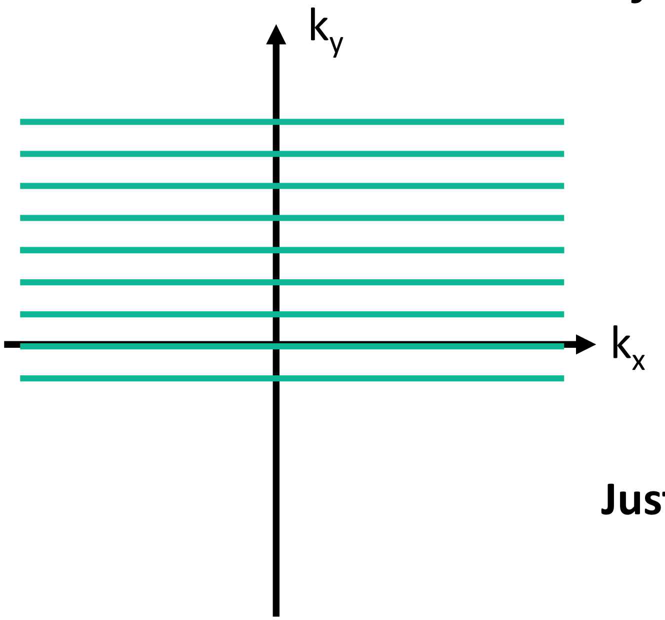


GRE or SE: one line of k-space per TR (usually 256, 512 lines)
Image time = $N_{phase} \times TR$

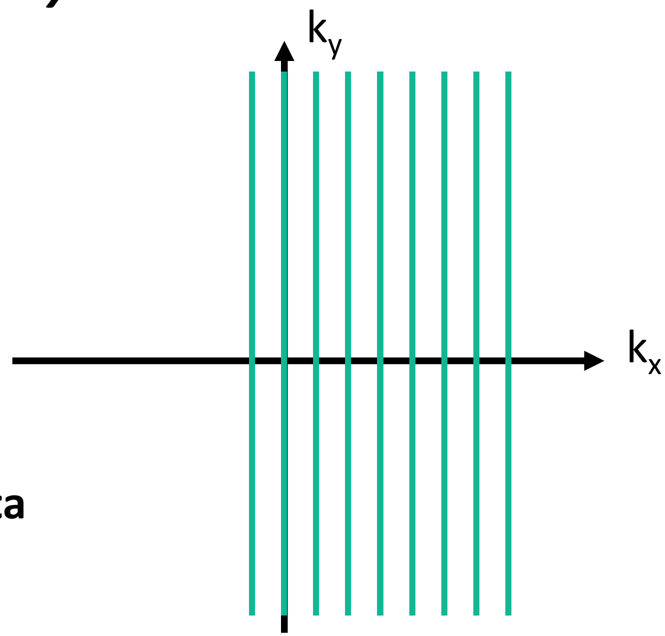
EPI: all lines of k-space per TR (typically 64 or 128)
Image time = TR

k-space trajectories

These techniques acquire part of k-space and ‘fill-in’ the rest using ***conjugate symmetry***



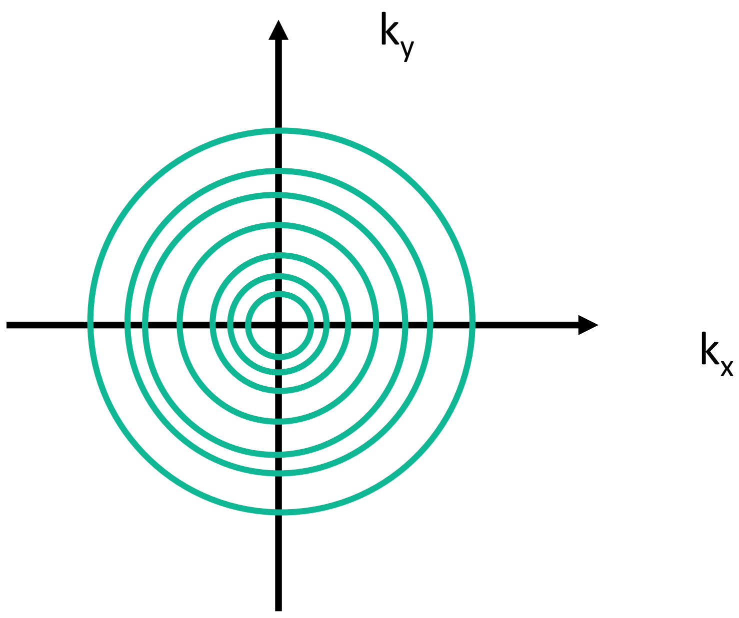
Just over half data collected

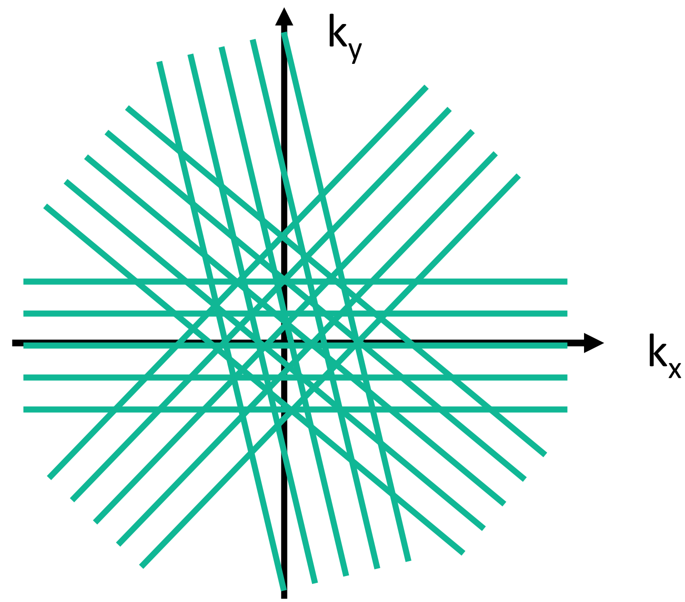


Partial Fourier (phase-conjugate symmetry / fractional NEX):
Collects half of phase-encode steps and speeds up imaging

Partial Echo (Read-conjugate symmetry):
Collects half of echo reducing the shortest possible TE

k-space trajectories





Spiral:
more efficient single shot
Centre of k-space is filled first

Radial:
Centre oversampled
Motion compensation

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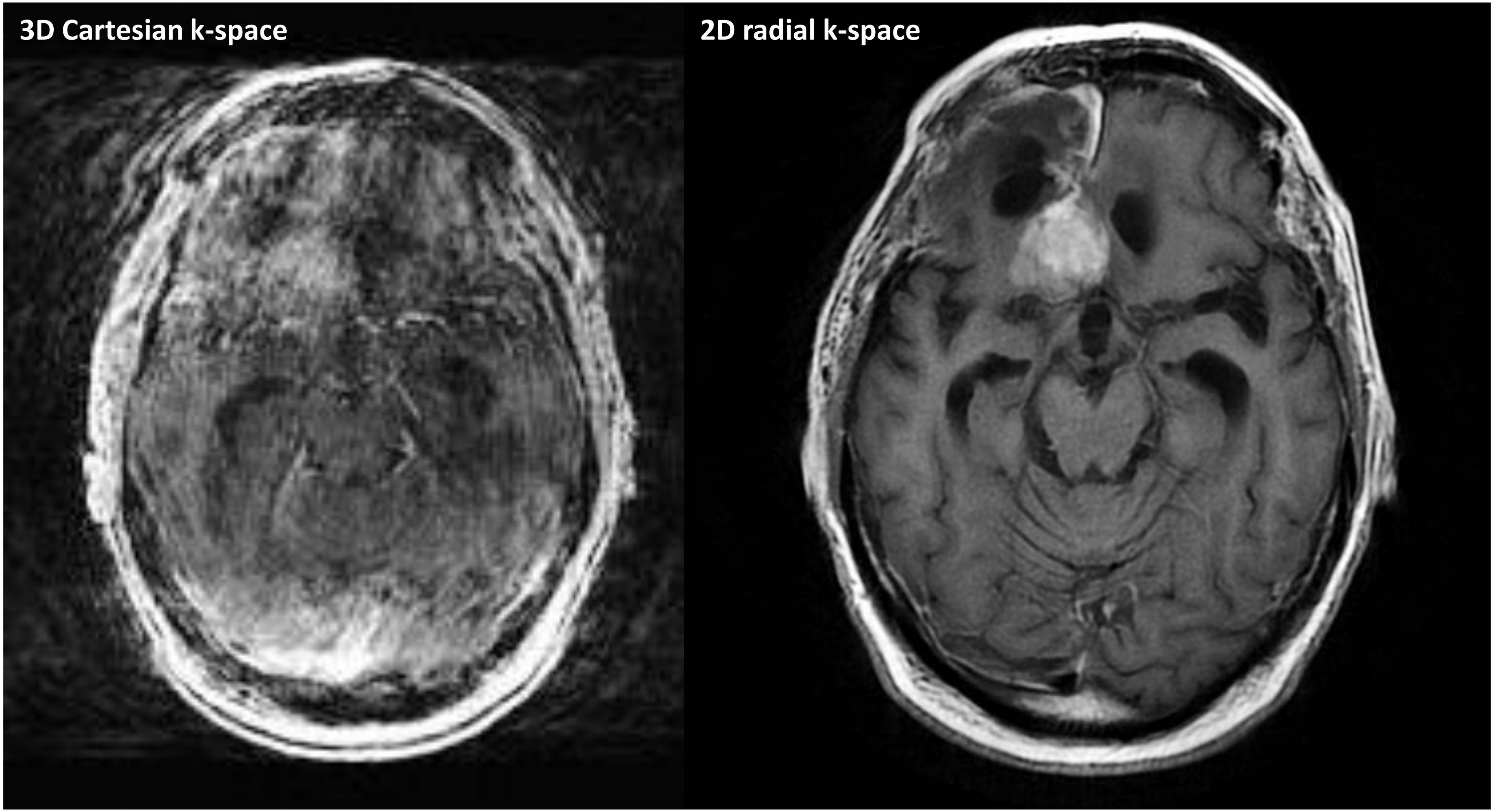
PROPELLER/BLADE/Multivane (radial k-space filling strategies)



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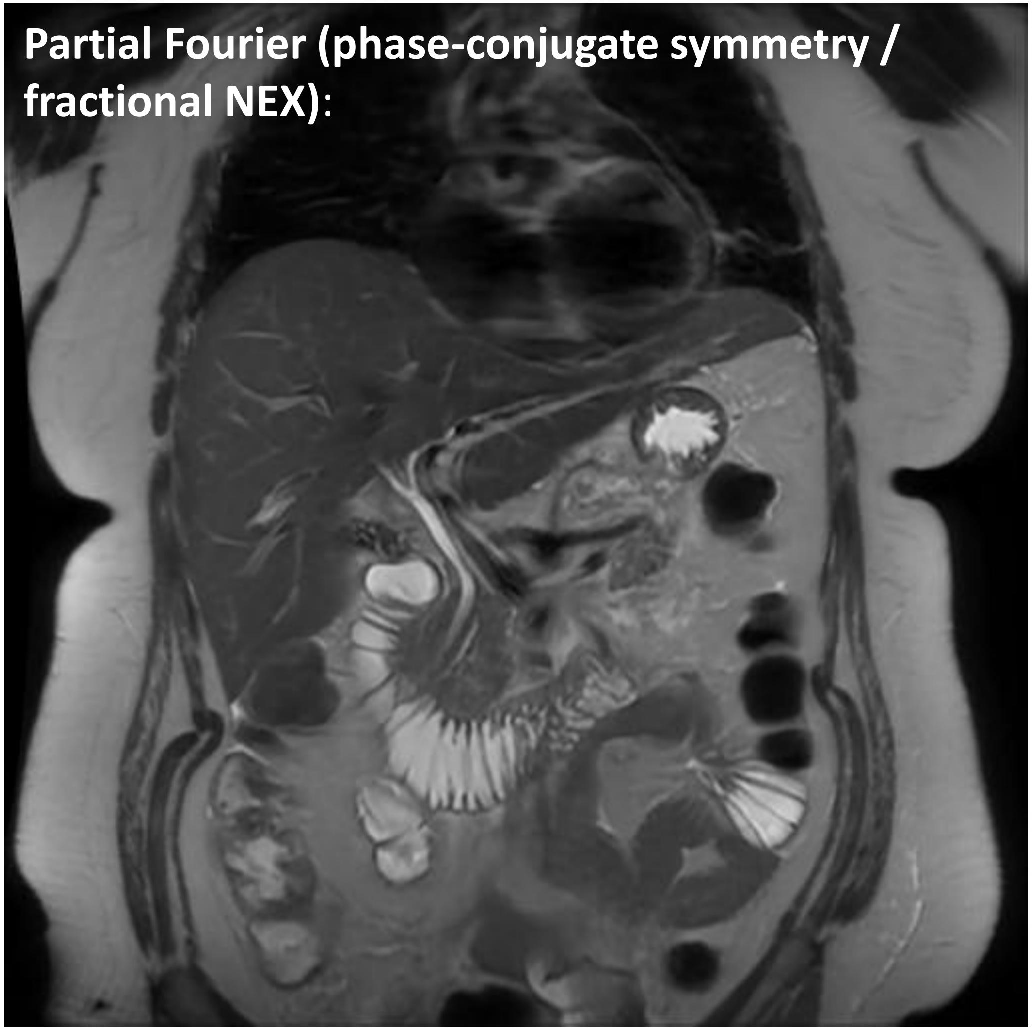
Partial k-space

- HASTE (Half Fourier Acquisition Single Shot Turbo Spin Echo)
- SS-FSE (Single-shot fast spin echo)
- Both fast/turbo spin-echo techniques used for sequential acquisition of T_2-weighted images
- Often make use of partial Fourier to reduce breath-hold time
- Short acquisition time makes sequence motion insensitive



Awareness of different k-space trajectories and their advantages/disadvantages

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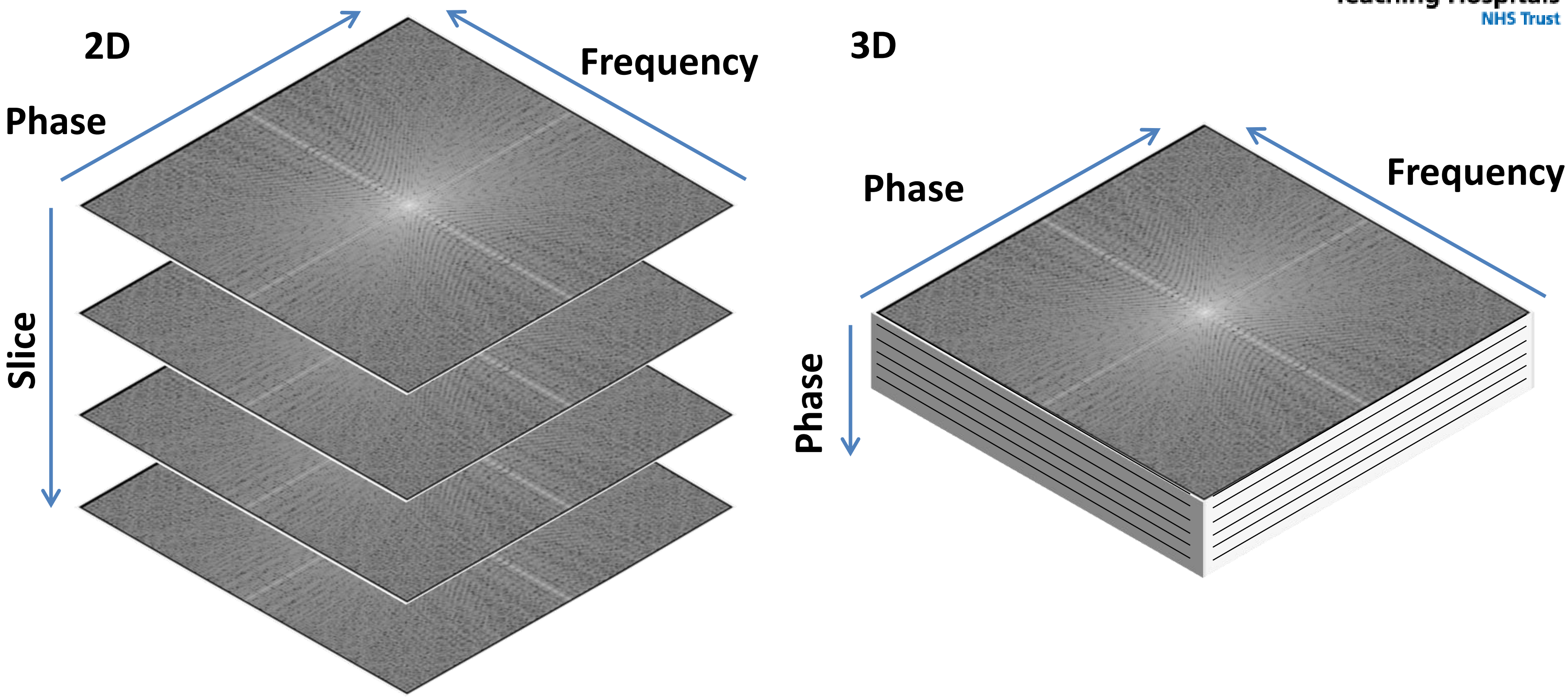
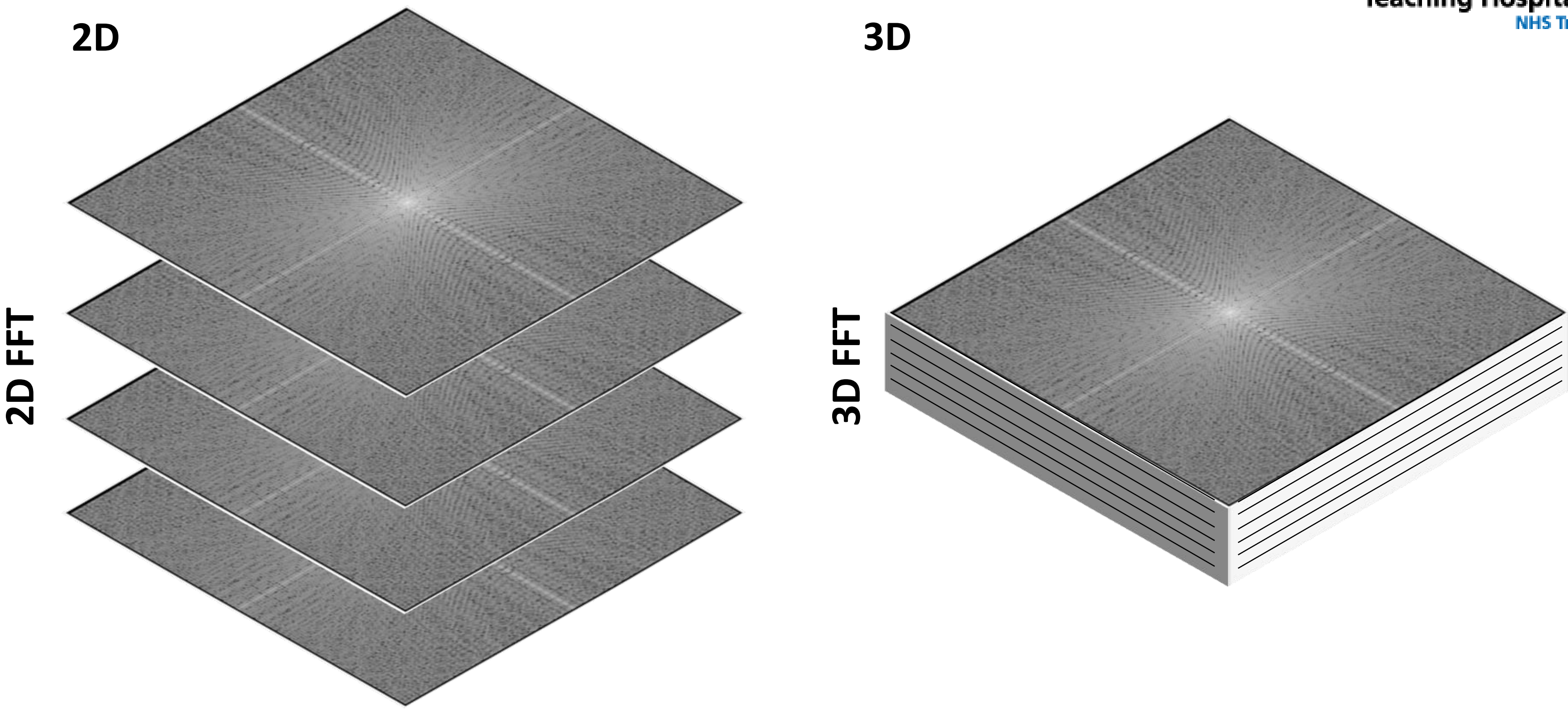
k-space trajectory	Advantages	Disadvantages
Cartesian	Simple to acquire. Minimal distortion artefacts. Works with parallel imaging	Prone to ghosting in PE direction. Requires complete filling of k-space. Image contrast generated ½ way through acquisition.
Spiral	Less prone to motion. Centre of k-space represents start of scan (dynamic scanning).	More affected by incorrect gradient timings. Image reconstruction is more complicated and require "gridding" of the data to warp the measured k-space points into a rectangular matrix for FFT.
Radial	Motion insensitive Generally high SNR	The corners of k-space are not sampled slightly reducing image sharpness. Complete coverage of the k-space circle (without gaps between the blades) takes ≈ 1.57 times as long than Cartesian filling method. FSE/TSE variants are SAR intensive.
Zig-zag	Very rapid filling of k-space (EPI) Single slice in 1 TR Motion Insensitive	Eddy currents lead to distortion. Long echo trains lead to low SNR. Generally poor near implants.
Partial Fourier (phase-conjugate symmetry)	Reduced acquisition time Preservation of spatial resolution	Reduced SNR. Square root % of data acquired. E.g. ½ data has 70% SNR compared to full k-space.
Partial Echo (Read-conjugate symmetry)	Shorter TE than normally possible. Shorter TE generates greater SNR. Good for time sensitive MR angiographic techniques.	Limited applications.

2D versus 3D sequences



- For 3D, excitation of the complete volume for each repetition
- Excitation repeated for each phase encoding in the 3rd Dimension
- Additional phase encoding direction also adds wrap in 3rd Dimension

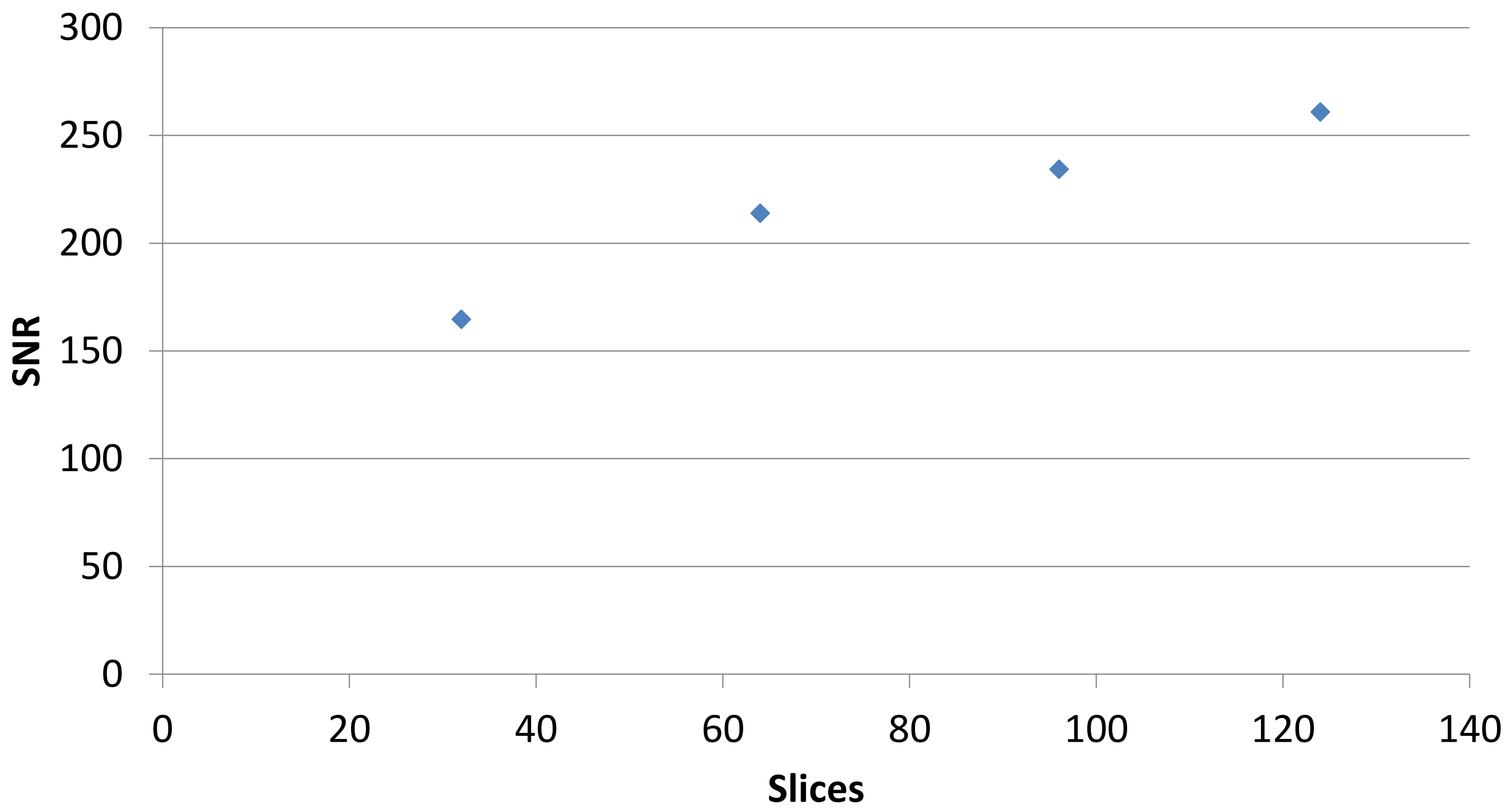
2D versus 3D sequences



- Every point in K-space goes to every point in image space
- 2D   x and y
- 3D   x, y and z!

2D versus 3D sequences

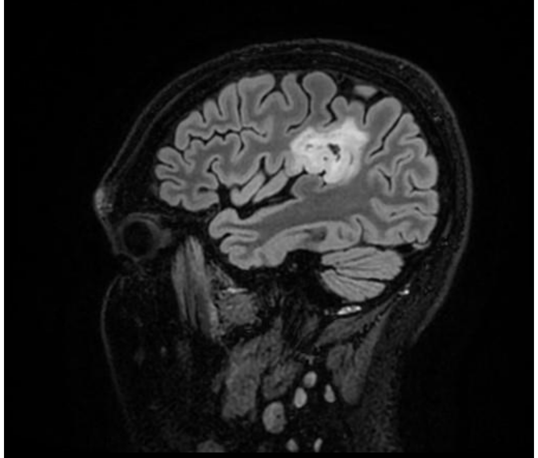
- For 3D sequences, SNR is proportional to the number of slices
- For 2D sequences, SNR is independent.



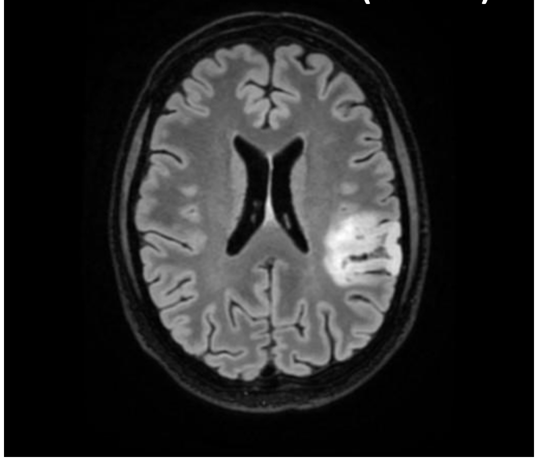
2D versus 3D sequences

- Motion at any point during a 3D acquisition will appear on all slices of a 3D dataset (vs only effected slices for 2D)
- 3D datasets have 2 phase encoding directions for wrap to occur (vs 1 for 2D)
- 3D sequences are more SNR efficient and thus allow thinner slices to be acquired (and enabling MPRs). Signal proportional to number of slices.
- Parallel imaging is allowed in 2 directions for 3D acquisitions occur (vs 1 for 2D)
- The thicker 2D slices often allow for higher in-plane matrix sizes compared to 3D.
- Flow artefacts are often improved on 3D sequences

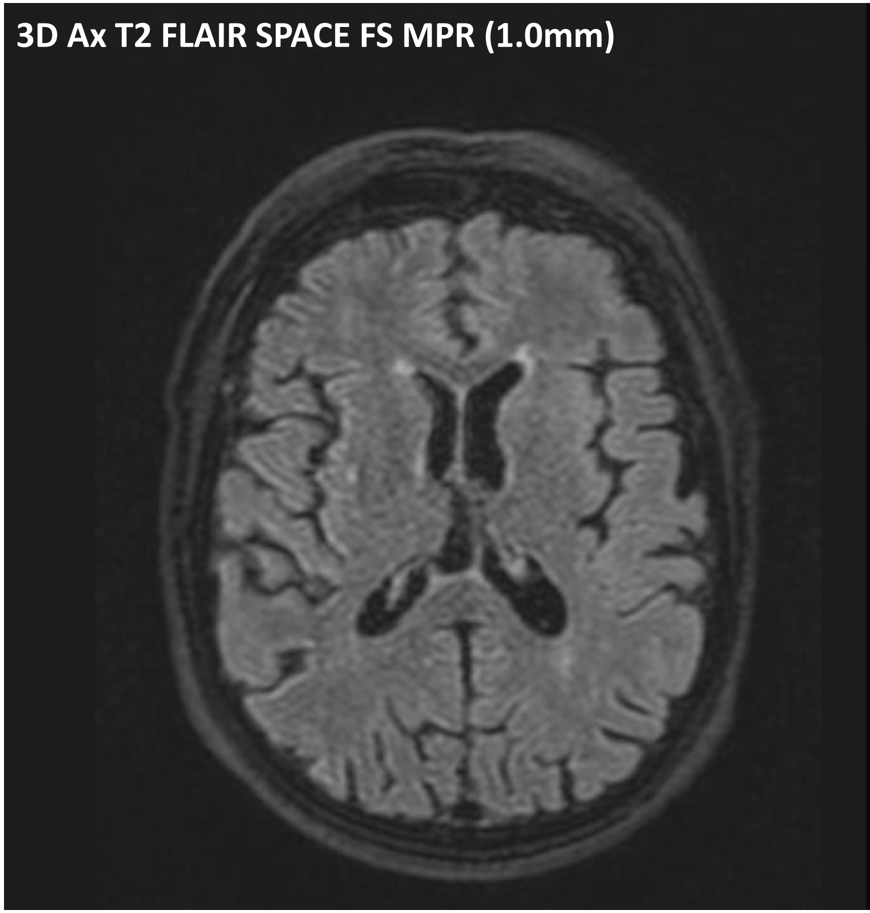
3D Sag 3D T2 FLAIR FS (0.5mm)



Ax T2 FLAIR FS MPR (1.0mm)



2D versus 3D sequences



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7.1 Creation, detection and spatial localisation of MR signal

- Nuclear magnetic resonance
  - *Nuclear spin (I). ^{1}H has two states. I=1/2. Principle of MRI*

- Precession about magnetic fields (B_0 and B_1)
  - *Spins will align with/against the main magnetic field (B_0). B_0 is typically 1.5 or 3.0T superconducting magnet. Rate of precession is related to B_0 (Larmor equation $\omega = \gamma B$). Can change align of spins with B_1 RF pulse. Bohr Condition - $B_1(t)$ must have components that rotate near the resonant frequency (ω_0)*

- Equilibrium magnetisation (M_0) and dependence on the strength of the magnetic field, B_0
  - *Net magnetic moment results from small difference in spin population. Increasing B_0 will increase population difference (Boltzmann distribution)*

7.1 Creation, detection and spatial localisation of MR signal

- Longitudinal (M_z) and transverse magnetisation (M_{xy})
  - *Net magnetic moment (net magnetisation) is 1 in M_z plane prior to B_1 excitation (RF). Following RF T_1 and T2 relaxation occur. M_{xy} shrinks. M_z grows. Simultaneous process. Block equations.*

- Slice Selection
  - *Magnetic field gradients. Can be any plane (double oblique). Applied simultaneously to RF pulses.*

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7.1 Creation, detection and spatial localisation of MR signal

- k-space:
  - Relationship between k-space and MR image
    - *Every point in k-space goes to every point in image space. Image contrast comes from centre of k-space. Fine detail is the rest of k-space.*
  - Frequency-encoding
    - *Magnetic field gradient used to create frequency difference across imaging plane. Occurs during echo formation and digitisation.*
  - Phase-Encoding
    - *Magnetic field gradient used to create phase difference across imaging plane. Different strength gradient according to k-space location. Slower process than FE.*
  - Awareness of different k-space trajectories and their advantages/disadvantages
    - *Conventionally line at a time. Can be optimised for time, artefact reduction or image contrast. Radial, spiral, Cartesian etc. Angios*

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7.1 Creation, detection and spatial localisation of MR signal

- 2D versus 3D sequences
  - *2D imaging uses multiple 2D inverse FFT. Movement artefacts localised to slices that motion occurred. 3D need all data acquiring to produce imaging. 3D inverse FFT. 3D artefacts appear on every image. 3D has more SNR and can produce thinner slices.*

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Question

- The outside of K-space contains spatial frequencies relating to?

    A. image resolution
    B. image contrast
    C. image detail
    D. image size

- Which type of coil produces more SNR at the surface of a patient?

    A. Volume Coil
    B. Phased Array Coil

Question

In magnetic resonance imaging the following are true:

A. Slice selection must always be applied to the z-axis

B. In a standard SE sequence, frequency encoding is applied during the signal acquisition

C. The steeper the slice selection gradient, the thinner the slice

D. In a standard SE sequence the image is built up line by line during acquisition

Question

In magnetic resonance imaging the following are true:

A. Prior to any radiofrequency (RF) pulse the total magnetic vector in the xy plane (m_{xy}) is equal to 1.

B. The signal measured in a 1.5T magnet will be proportionately more than a 3.0T magnet.

C. The net magnetic vector in the z-axis provides a recordable MR signal.

D. A standard 1.0T MRI scanner produces a magnetic field 100 time's that of the earth.

Question

In magnetic resonance imaging the following are true:

A. Slice selection is applied simultaneously with the initial RF excitation

B. A row of phase encoding data in K-space takes less time to acquire than a column of frequency encoding data

C. The centre of a K-space contains the data relating to high spatial resolution

D. Slice thickness can be reduce by decreasing the RF bandwidth for each slice (assuming the gradient remains the same)

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Question

In magnetic resonance imaging the following are true:

A. The main system magnet is normally a permanent magnet design

B. Superconductivity requires cooling near to 0°C

C. Gradients for spatial encoding are provided by altering the main magnet

D. The main magnetic field is normally define to be along the Z-direction

Question

Concerning T_1, which of the following are true:

A. It is the time taken for transverse recovery to reach 37% of the maximum value.

B. T_1 is always longer than T_2.

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Question

Concerning T_2, which of the following are true:

A. Occurs due to spin-lattice relaxation.

B. Is affected by magnetic field inhomogeneities.

C. When 63% of the transverse signal is lost, this is referred to as time T_2

Question

In magnetic resonance imaging the following are true:

A. FID is a commonly used basic sequence

B. A spin-echo sequence removes the dephasing effect of field inhomogeneities with a second 90° pulse

C. The longer the TE, the smaller the subsequent signal