NUCLEAR MAGNETIC RESONANCE IMAGING SYSTEM

- Fundamentals of nuclear magnetic resonance (NMR) –
  - Magnetic moment
  - Free induction decay (FID)
  - Fourier transformation of FID
- Generation and detection of NMR signal –
  - The magnet
  - Room temperature magnetic gradient
  - NMR coil/probe
  - Transmitter
  - Receiver
- Image reconstruction techniques –
  - introduction
  - Sequential point method
  - Sequential plane method
  - Discrimination based on relaxation rates
- Biological effects of NMR imaging
- Basics of Magnetic Resonance Imaging
- Fundamentals of nuclear magnetic resonance
- Introduction to MRI sub systems

INTRODUCTION

Interaction of radio frequency field and a locally variable static magnetic field to view anatomical structures and composition

- Imaging is done with magnetic fields, so there is no ionizing radiation
- Flexible imaging - 2D, 3D, cine, real-time
- More control of imaging plane than other modalities
- Safety profile similar to ultrasound
- Excellent tissue contrast - multiple contrast mechanisms
- Molecular imaging - based on intrinsic properties of molecules and their surrounding environment
- Superior tissue contrast to other modalities

HISTORY

Nuclear Magnetic Resonance

1940s – 1950s

Edward Purcell

Felix Bloch

- Independently discovered the nuclear magnetic resonance effect
- Awarded Nobel Prize in 1952 for their discoveries

[1] Tal Geva JCMR 2006

1970s : The birth …

- 1971: Raymond Damadian – relaxation times for tumors in rat models of cancer
- 1973: Abe Zenuemon and others, file patent for the first targeted NMR for evaluation of information inside from outside
- 1974: Magnetic Resonance Imaging is born by the use of gradients, thanks to Paul Lauterbur and Peter Mansfield
- 1975: Richard Ernst (Anil Kumar et. al) described the use of FT for MRI

[1] Tal Geva JCMR 2006

*Raymond Damadian formed the first company **FONAR** to manufacture MR scanners for clinical use in 1982*

ERA OF MRI

- Prof. Peter Mansfield was awarded Nobel in 2003 for his discoveries in MRI (with Prof. Paul C . Lauterbur of USA)
- Peter Mansfield is from Nottingham University, UK

OVERVIEW OF MRI SYSTEM

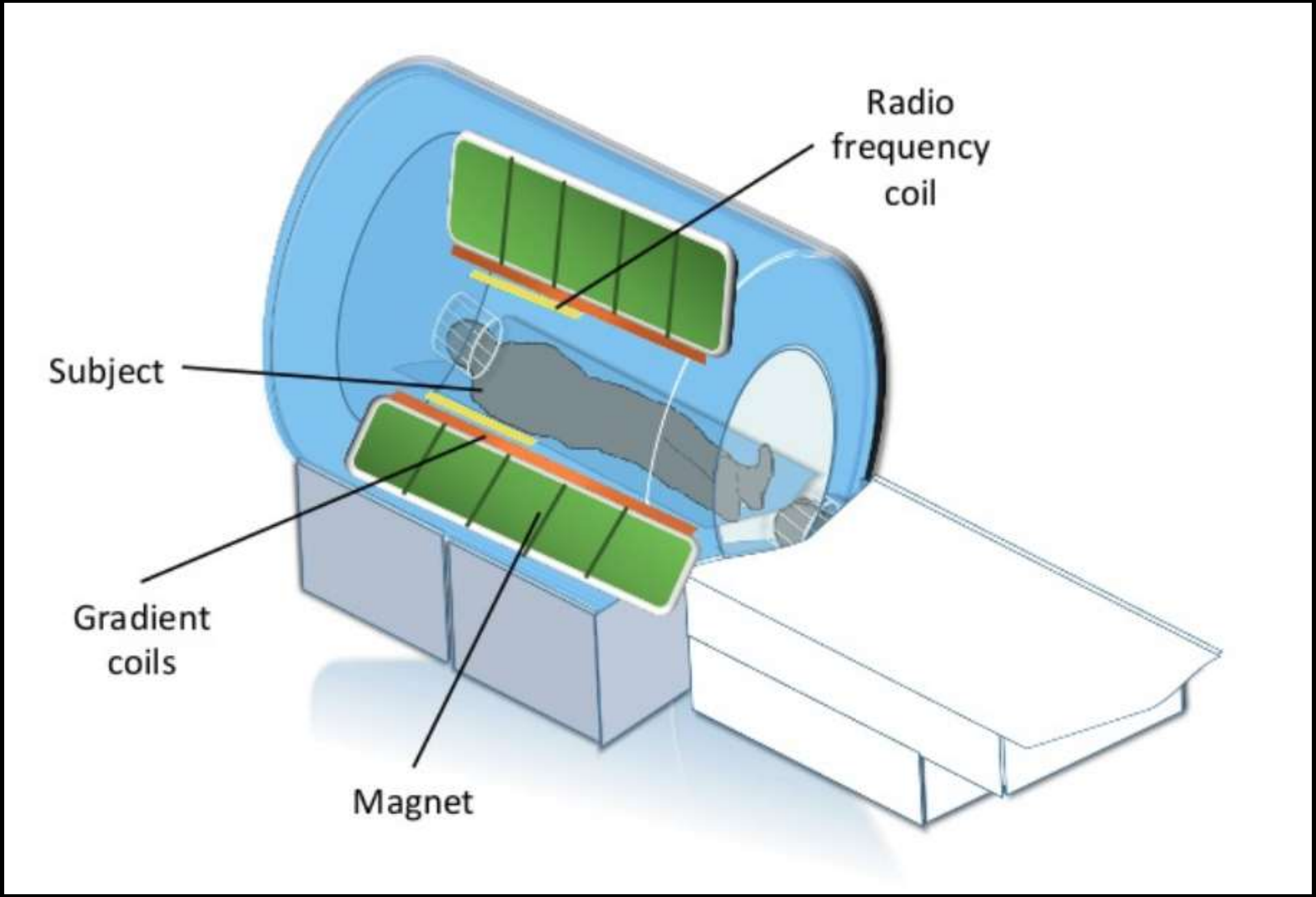
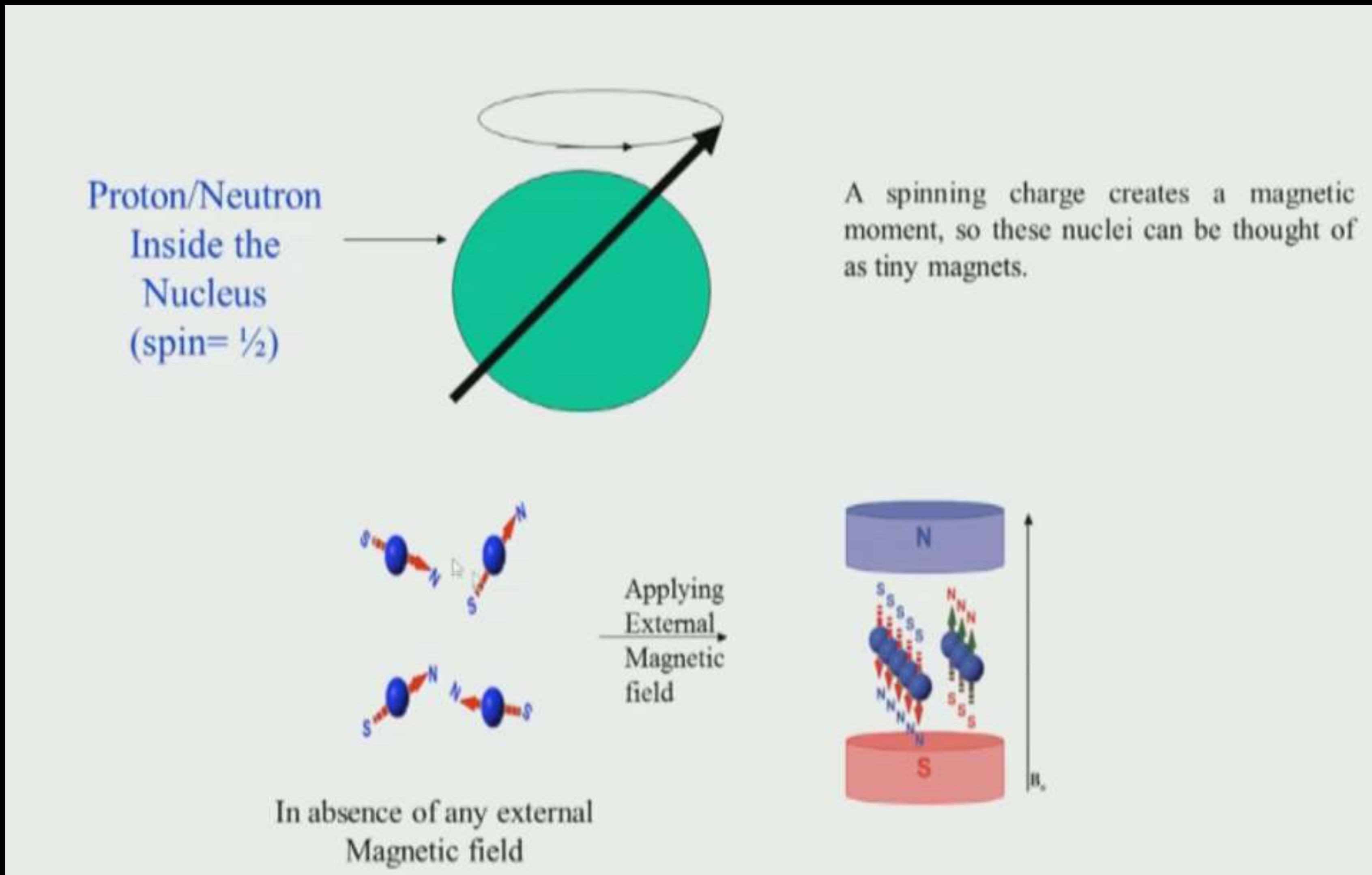


Image courtesy: MRI scanner cuaway: Colinmcnulty.com

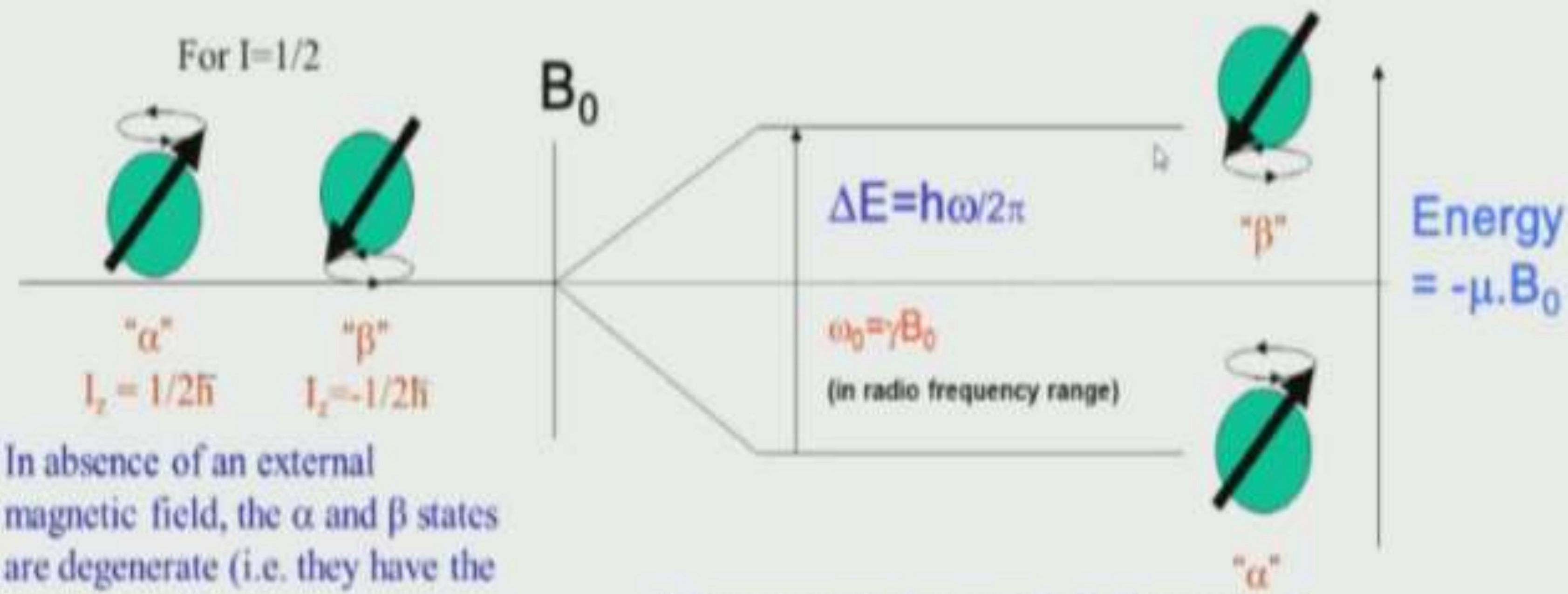
MAGNETIC MOMENT



Quantum Mechanical Picture

- The system is thus split into two states with different energies:

$E_{1/2} = -1/2\ h\ B_0$ and $E_{-1/2} = 1/2\ h\ B_0$ => $\Delta E = E_{-1/2} - E_{1/2} = hB_0$

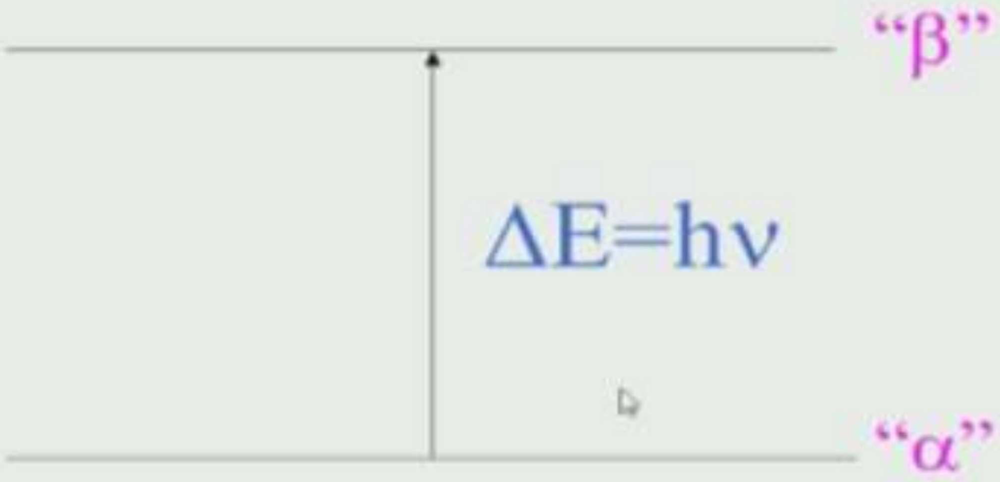


In absence of an external magnetic field, the α and β states are degenerate (i.e. they have the same energy)

When an external magnetic field is applied these two states split into two distinct levels

Quantum Mechanical Picture

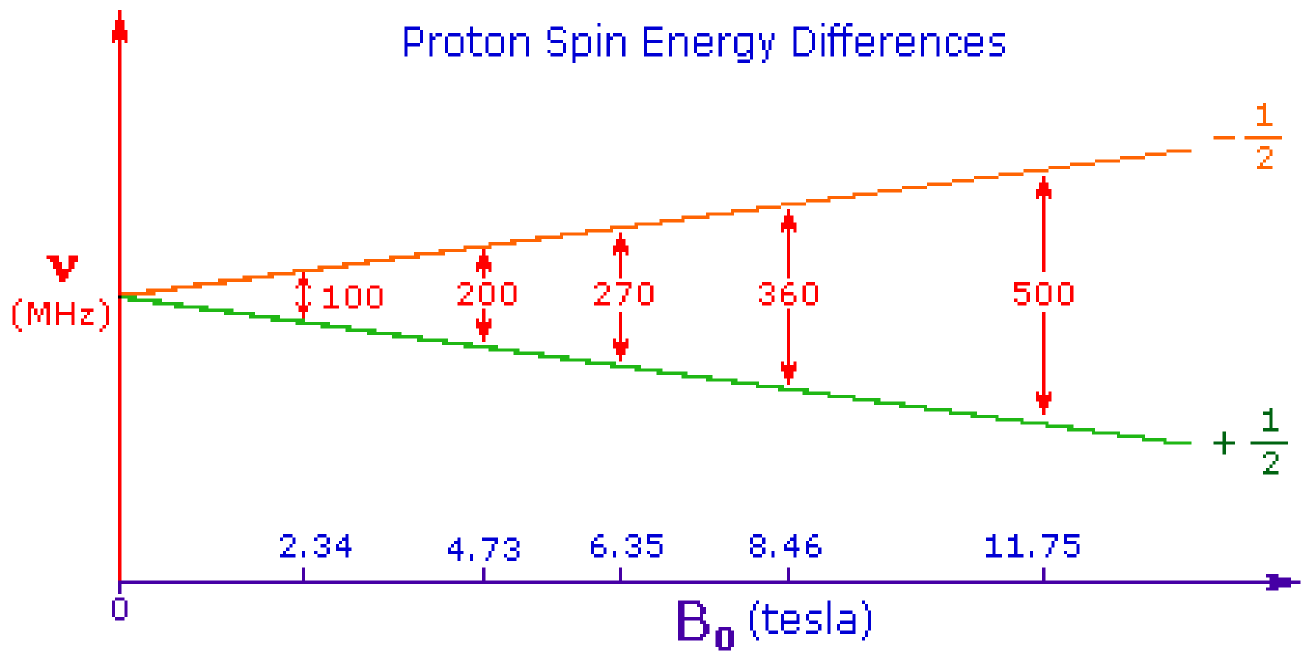
- We can induce transition from 'α' to 'β' state or vice-versa by supplying energy *equal* to the gap between the two states



- Hence the name 'nuclear magnetic resonance' (resonance results in energy transfer)
- The number of nuclei in the ground state is more than in the excited state at equilibrium. Hence, there is a net absorption. The strength of the signal is proportional to the net difference in population between the two states.

NUCLEAR MAGNETIC RESONANCE

- Field-frequency correlation for ^{1}H-nucleus
  - ✓ Magnetic field: ~ 10 000 oersted
  - ✓ Resonance frequency: ~ 42.5 MHz



NUCLEAR MAGNETIC RESONANCE

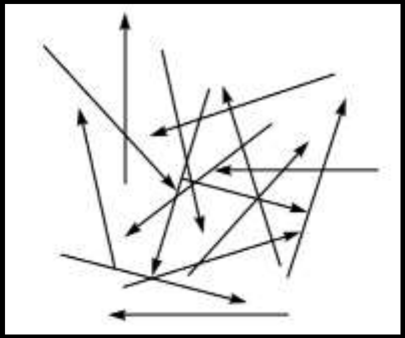
- When a nucleus with a magnetic moment is placed in a magnetic field, the energy of the nucleus is split into lower (moment parallel with the field) and higher (anti-parallel) energy levels.
- The energy difference is such that a proton with specific frequency (energy) is necessary to excite a nucleus from the lower to the higher state. The excitation energy E is given by the Planck's equation

$$E = h\omega_o$$

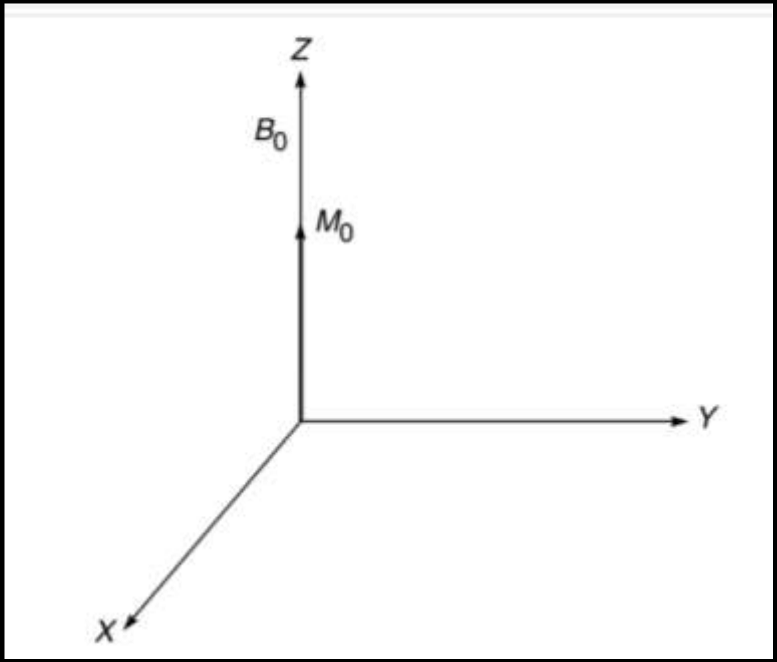
Where h is Planck's constant divided by 2π (h – 6.63×10^{-34} J-sec)

This energy is usually supplied by an RF magnetic field

MR physics

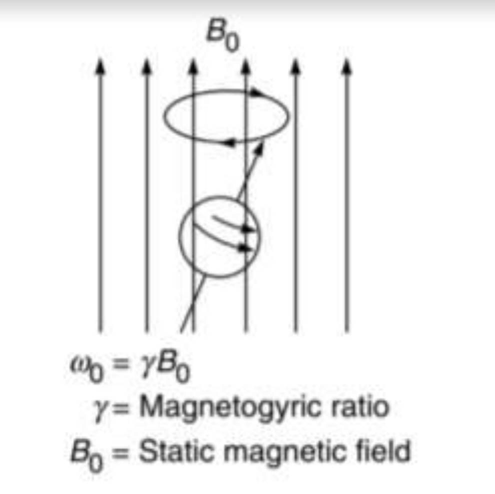


Random alignment of magnetic moments of the nuclei making up the tissue, resulting in a zero net magnetisation.



The application of external magnetic field causes the nuclear magnetic moments to align themselves, producing a net moment in the direction of the field B_0

MR physics

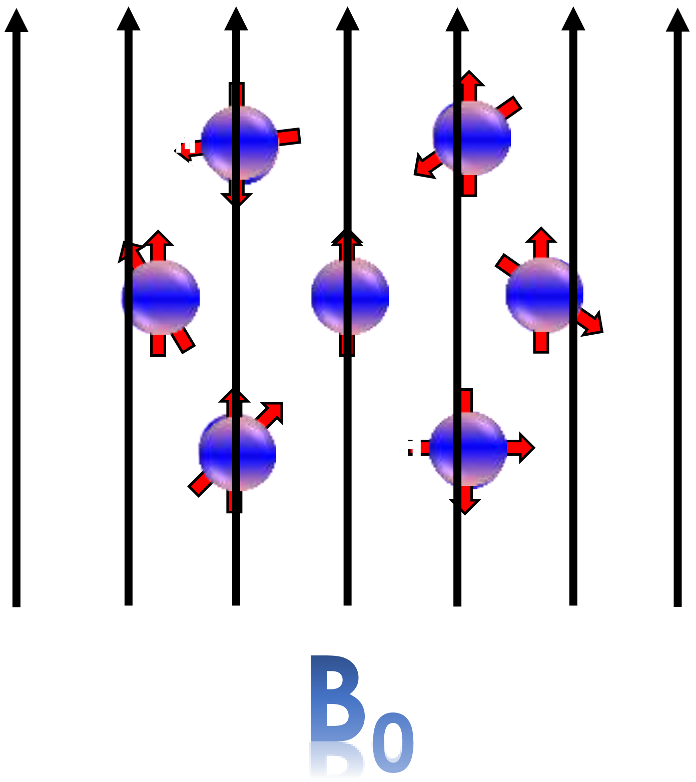


Precessing or wobbling of the nucleus about an applied magnetic field, with a resonant angular frequency ω_0

Nuclei	*Nuclear spin*	*$\gamma/2\pi$ (MHz/kg)*
^1H	1/2	4.26
^2H	1	0.65
13C	1/2	1.07
15N	1	0.31
19F	1/2	4.01
31P	1/2	1.72

MR physics

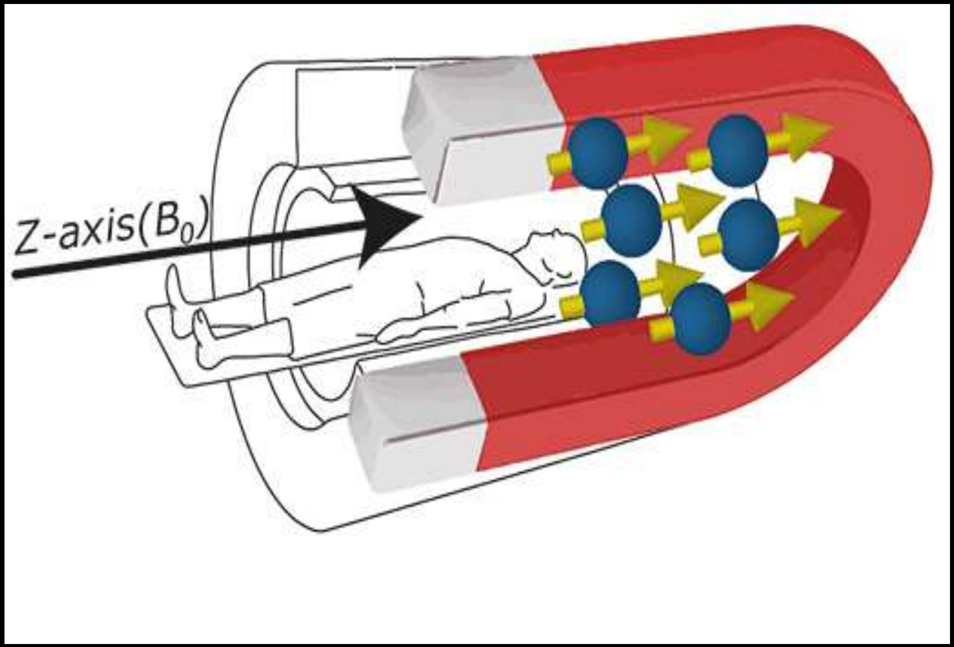
FID

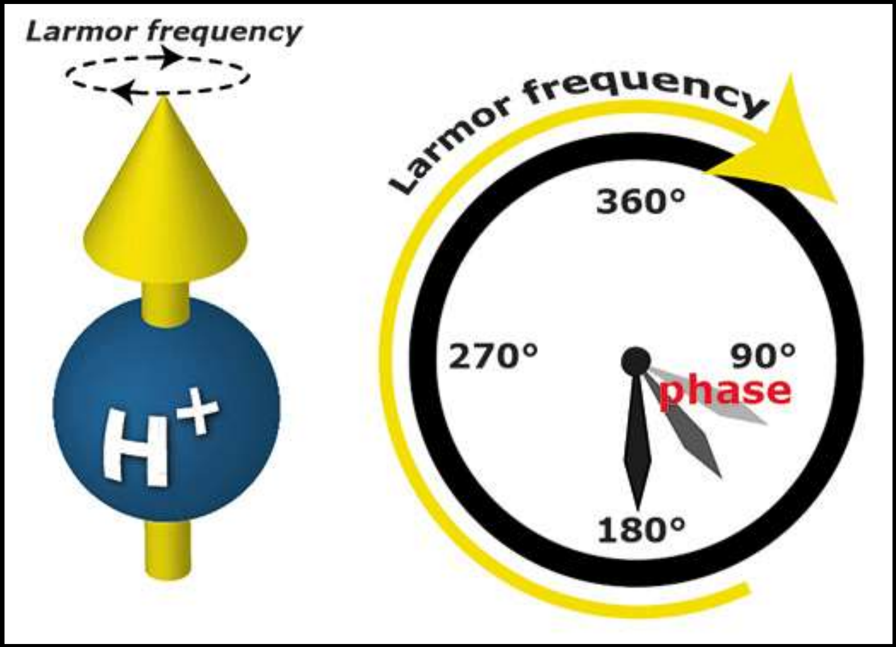


MAGNETIC MOMENT

- Nuclei containing an odd number of protons or neutrons or both in combination, possess a nuclear ‘spin’ and a magnetic moment which has both magnitude and direction
- In body tissue or any other specimen, the magnetic moments of the nuclei making up the tissue are randomly aligned and have zero net magnetization ($M = 0$)
- The application of external magnetic field causes the nuclear magnetic moments to align themselves, producing a net moment ($M0$)in the direction of the field $B0$

PRECESSION


Z-axis(B_0)


Larmor frequency
Larmor frequency
H^+
360°
270°
90°
phase
180°

PRECESSION

- Hydrogen proton which possesses a magnetic moment attempts to align itself with the magnetic field in which it is placed
- Results in a precession or wobbling of the magnetic moment about the applied magnetic field with a resonant angular frequency, ω_o (called the Larmor frequency) are determined by a constant γ (the gyromagnetic ratio) and the strength of the applied magnetic field B_0. Each nuclide possesses a characteristic value for γ but ω_o and B_0 are related as in Larmor equation.

Resonance frequency

$$\omega_0 = \gamma B_0$$

Or, when gradient present,

$$\omega_0 = \gamma (B_0 + G_r r)$$

RESONANT FREQUENCY AT 1T

- ^{1}H- 42.57MHz
- ^{19}F- 40.05MHz
- ^{31}P – 17.24MHz
- ^{13}C -10.71MHz
- ^{23}Na – 11.26MHz

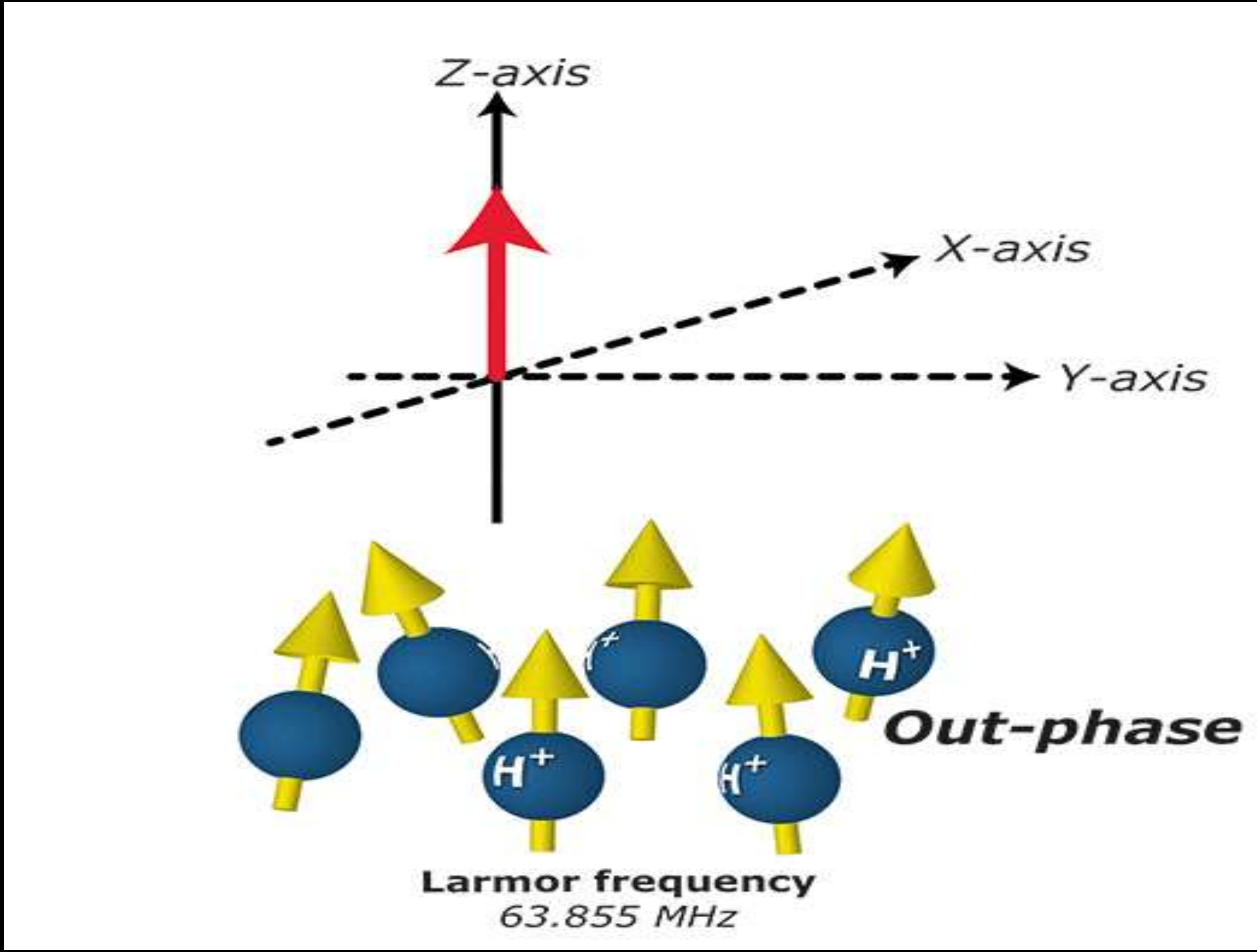
• Table 22.2 ***Characteristics of Common Biological Isotopes***

Element	*% of body weight*	*Isotope*	*NMR frequency MHz/T*
Hydrogen	10	^{1}H	42.57
Carbon	18	^{13}C	10.70
Nitrogen	3.4	^{14}N	03.08
Sodium	0.18	^{23}Na	11.26
Phosphorus	1.2	^{31}P	17.24

MAGNETIC FIELD STRENGTH

- Gauss or Tesla is unit of measurement.
- 1 gauss = 10^{-4}T
- The strength of the field at the Earth's surface ranges from less than 30 microteslas (0.3 gauss) in an area including most of South America and South Africa to over 60 microteslas (0.6 gauss) around the magnetic poles in northern Canada and south of Australia, and in part of Siberia.
- 1.5T and 3T MRI machines are widely used clinical scanners.

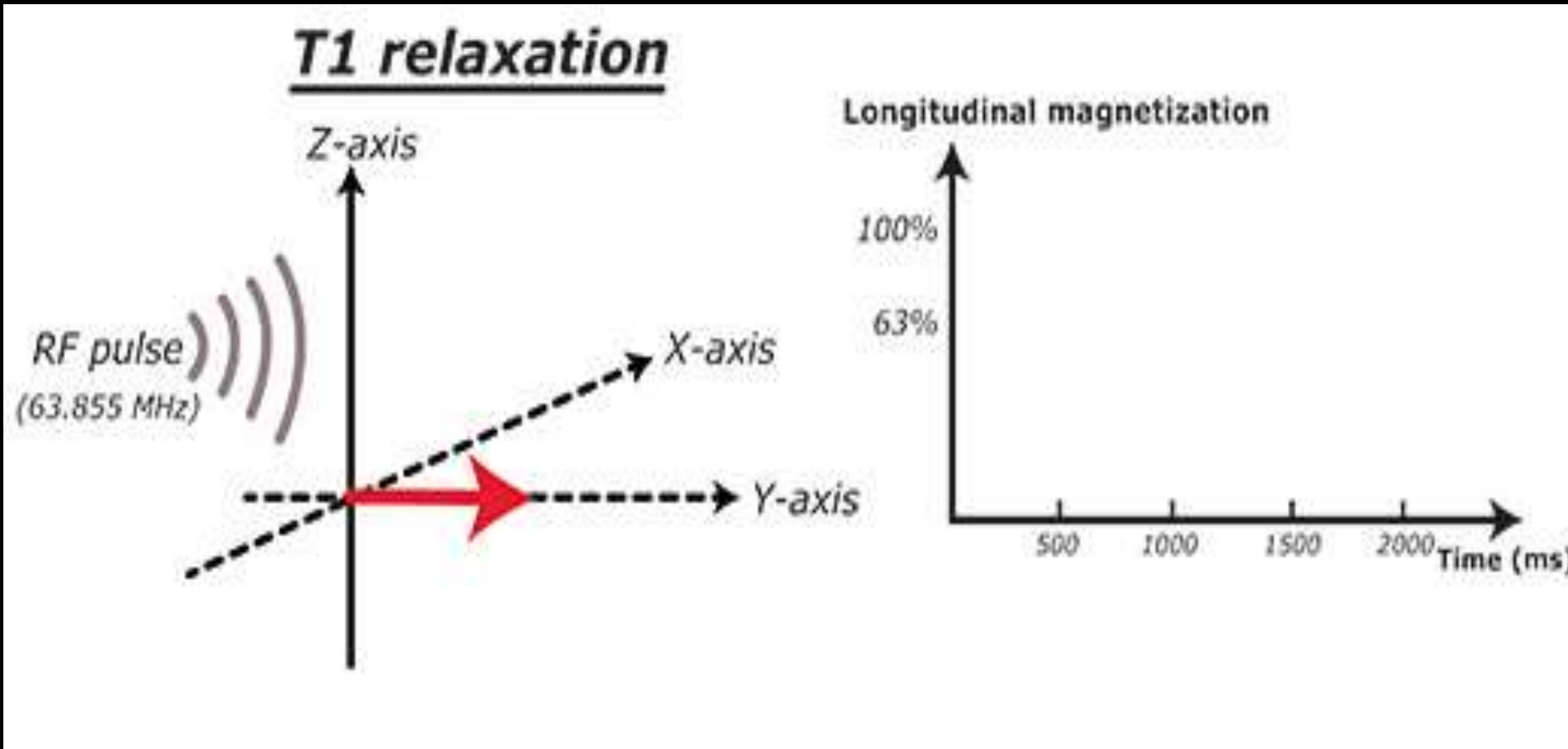
EXCITATION



RELAXATION TIMES

- Each tissue is characterized by two relaxations times: T1 (longitudinal relaxation time) and T2 (transverse relaxation time)
- Most of the images are created by one of these two characteristics being the predominant source of contrast
- This means when an image is described as a T1-weighted image T1 is the main source of contrast
- Different tissues have different relaxation times. These relaxation time differences is used to generate image contrast

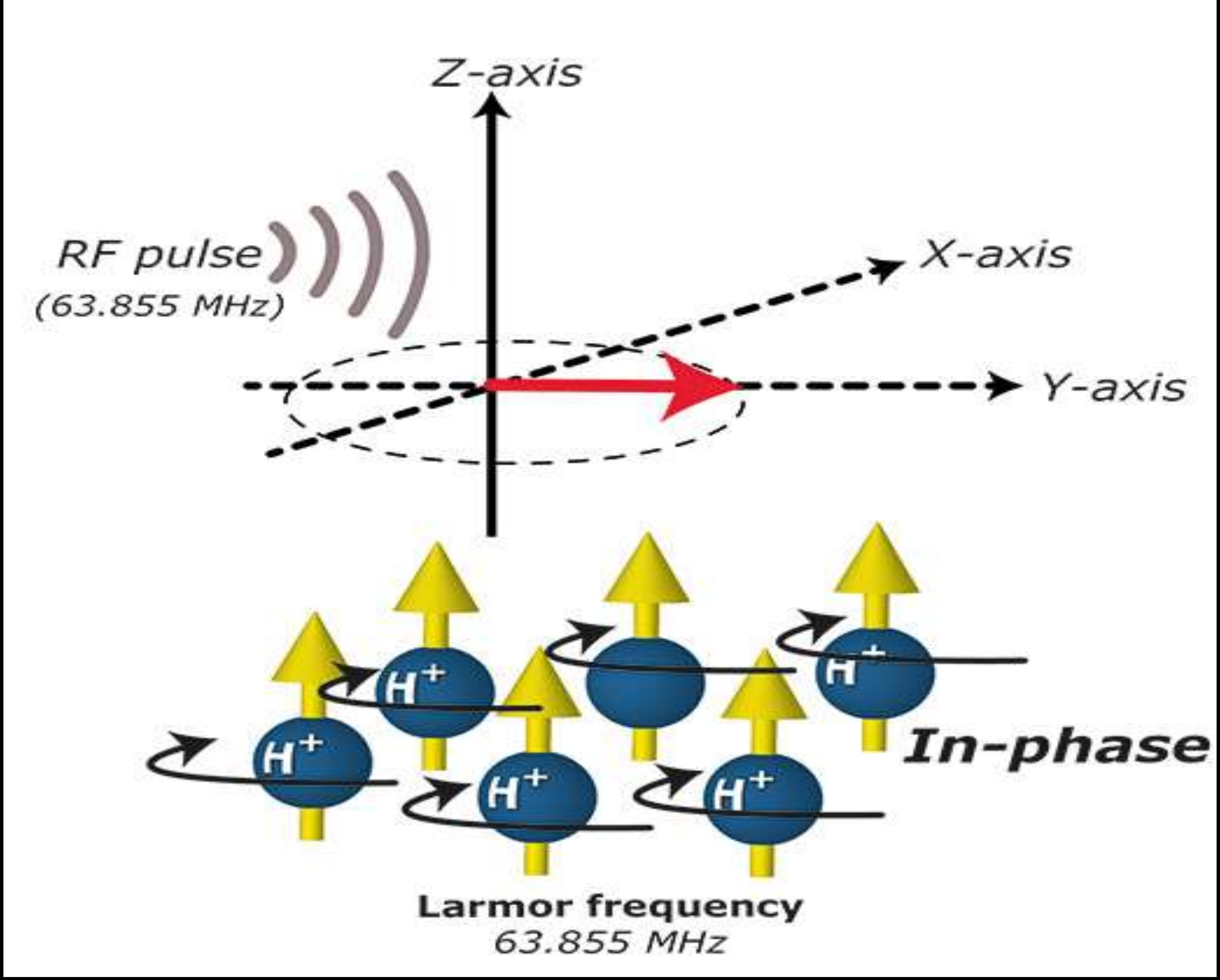
T_1 relaxation



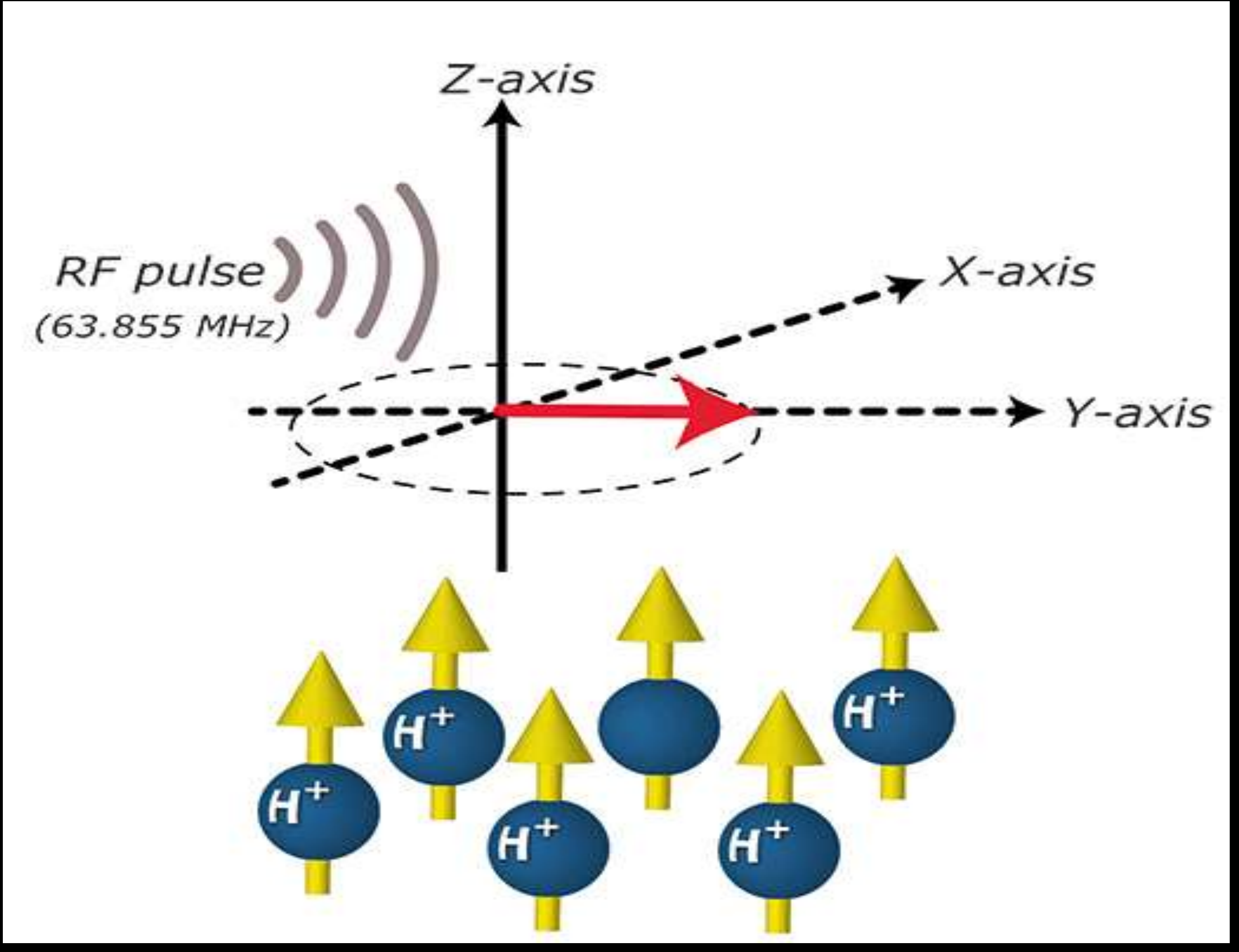
NMR Frequency in MHz/T

- Hydrogen – 42.57
- Carbon – 10.70
- Nitrogen- 3.08
- Sodium – 11.26
- Phosphorus – 17.24

T_2 relaxation

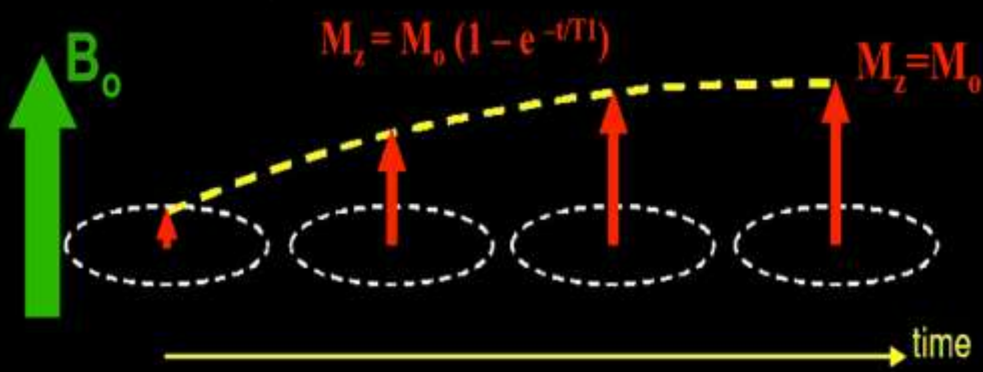


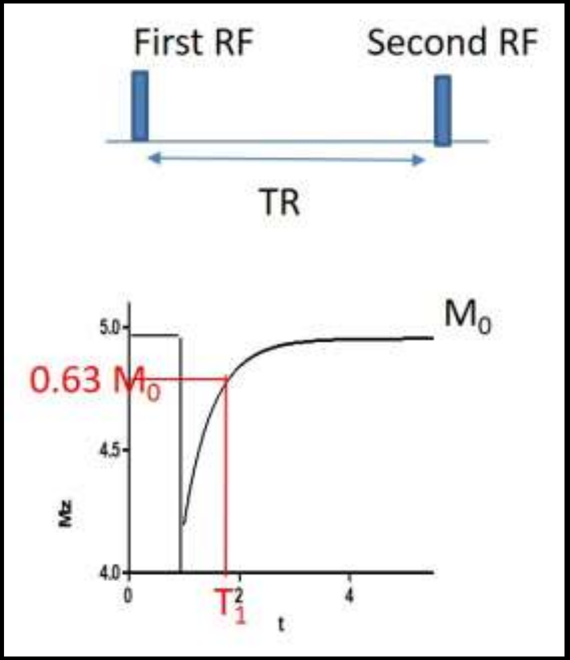
T_2 relaxation



T_1 RECOVERY

- Spin lattice interaction result in re-growth of longitudinal magnetization
- Rate of change of longitudinal magnetization is captured by an exponential recovery, is a cross product of magnetization moment M and the applied external field B
- Synonyms: longitudinal relaxation, thermal relaxation and spin-lattice relaxation

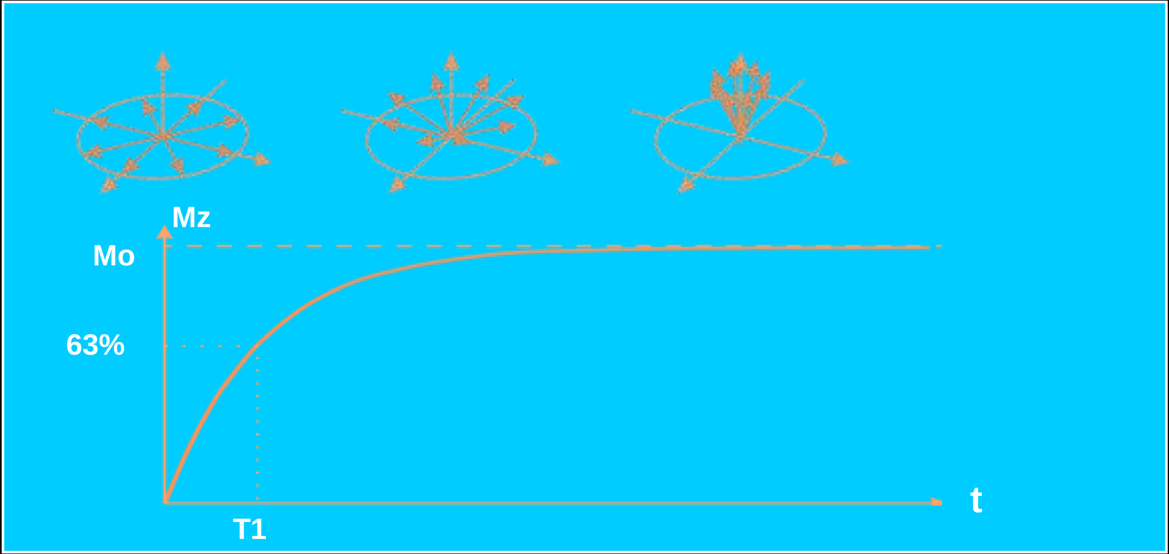




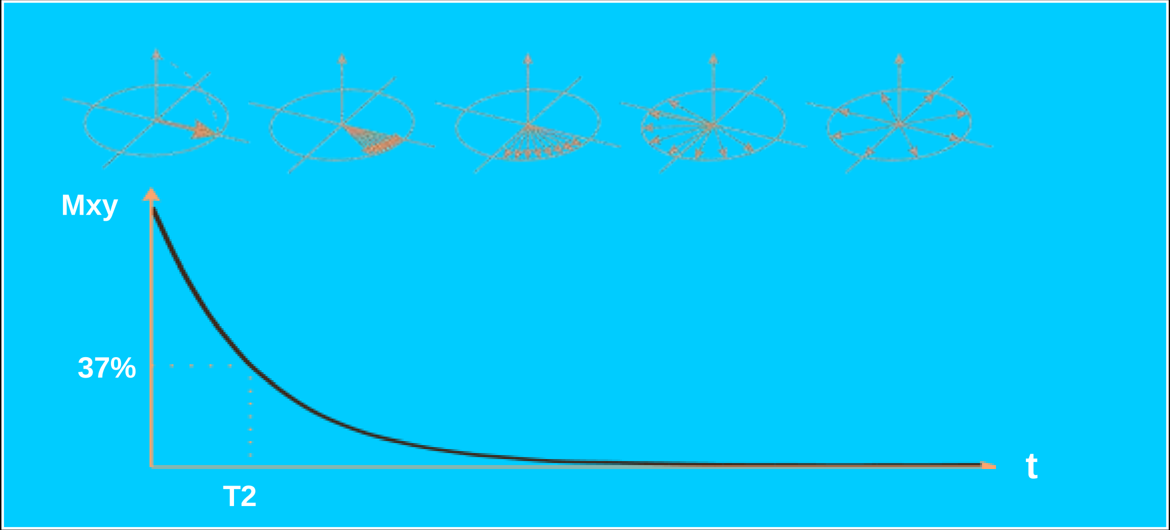
http://mriquestions.com
Magnetic resonance imaging: Physical principles and sequence design

Relaxation time

Longitudinal Relaxation Time T_1



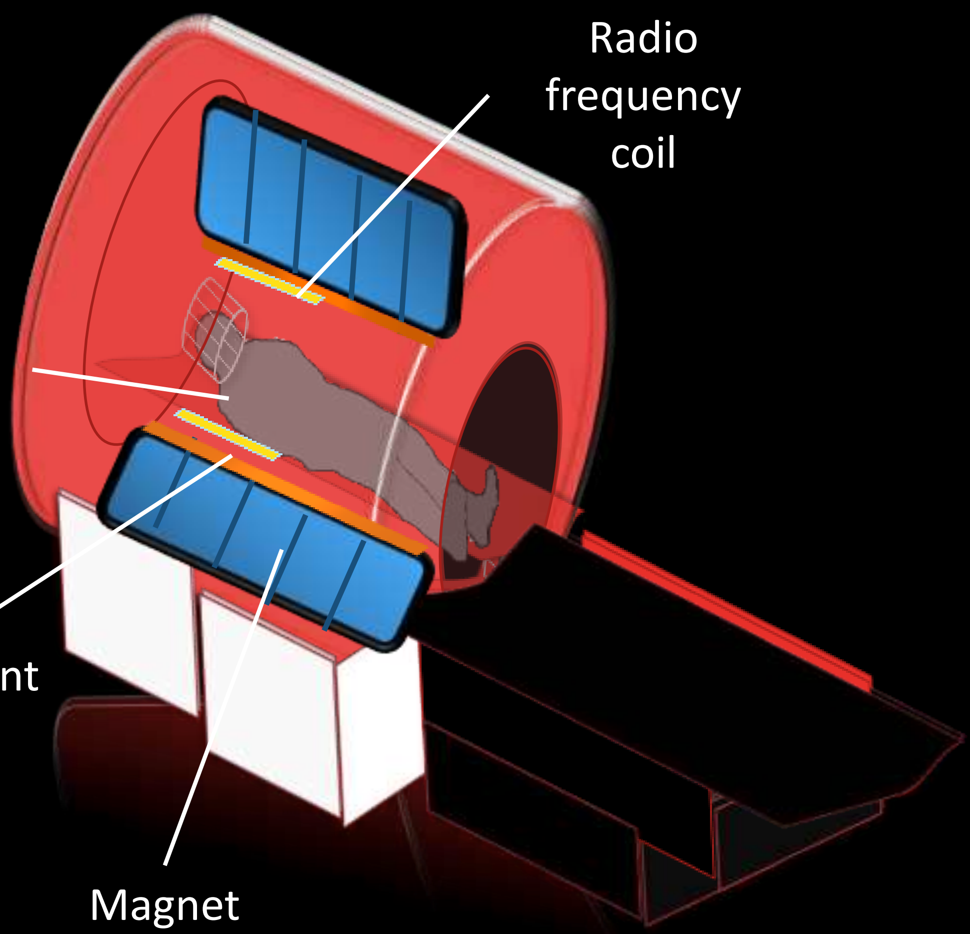
Transverse Relaxation Time T_2



T1 and T2 relaxation time

- Constant T_1 (spin-lattice time) governs the evolution of Mz towards its equilibrium value Mo and involves the dissipation of energy from collection of nuclei, the spin system to the atomic and molecular environment of nucleus ‘the lattice’
- Constant T_2 (spin-spin relaxation time) governs the evolution of the magnitude of the transverse magnetization, $M_x i+M_y j$, towards its equilibrium value of zero.

Overview of a MRI system



Representation of a typical MRI scanner

Image courtesy: MRI scanner cutaway: MIRC

MRI PRINCIPLE

- Hydrogen nucleus has an unpaired proton(paramagnetic) which is positively charged
- Every hydrogen nucleus is a tiny magnet which produces small but noticeable magnetic field
- Hydrogen atom is the only major species in the body that is MR sensitive
- Hydrogen is abundant in the body in the form of water and fat
- Essentially all MRI is hydrogen (proton) imaging

MRI PRINCIPLE

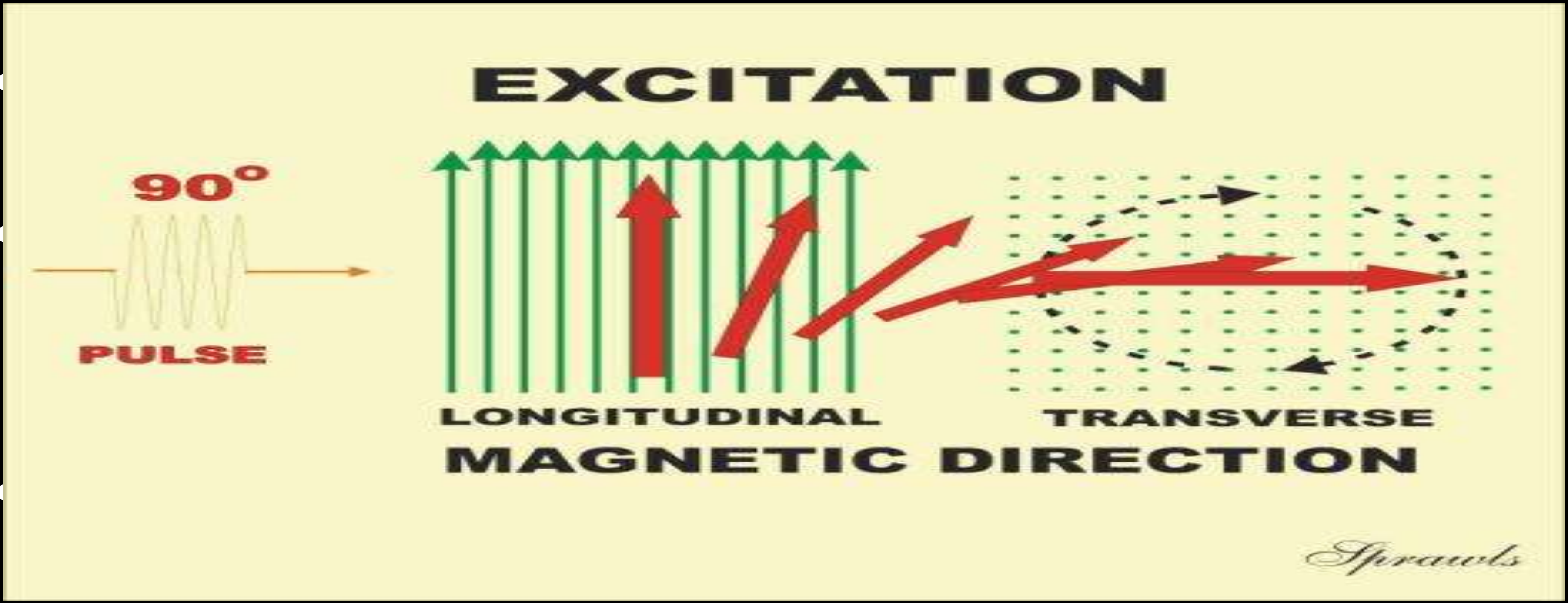
- In an MR experiment there are three magnetic fields
- B_0- Static magnetic field created by the superconducting magnet and the gradient coils
- The B_0 field by itself needs to be as uniform as possible usually achieved by shimming
- The Gradient Field modify the B_0 field in kilo hertz
- B_1 is the magnetic field created by the RF excitation
- B_1 usually changes in hundred‘s of mega hertz

BODY IN AN EXTERNAL MAGNETIC FIELD (B_0)

- Body has many such atoms that can act as good MR nuclei (^{1}H, ^{13}C, ^{19}F, ^{23}Na)
- Hydrogen nuclei is one of them which is not only positively charged, but also has magnetic spin
- MRI utilizes this magnetic spin property of protons of hydrogen to elicit images
- When a patient is placed in the magnet the hydrogen atoms in the water of their body tissues line up along the magnetic field.
- Radiofrequency pulses are sent in, causing the atoms to ‘flip’ into another plane and then ‘relax’ back when the pulse is turned off. This recovery process is known as ***relaxation*****.**
- This relaxation time varies from one type of tissue to another.

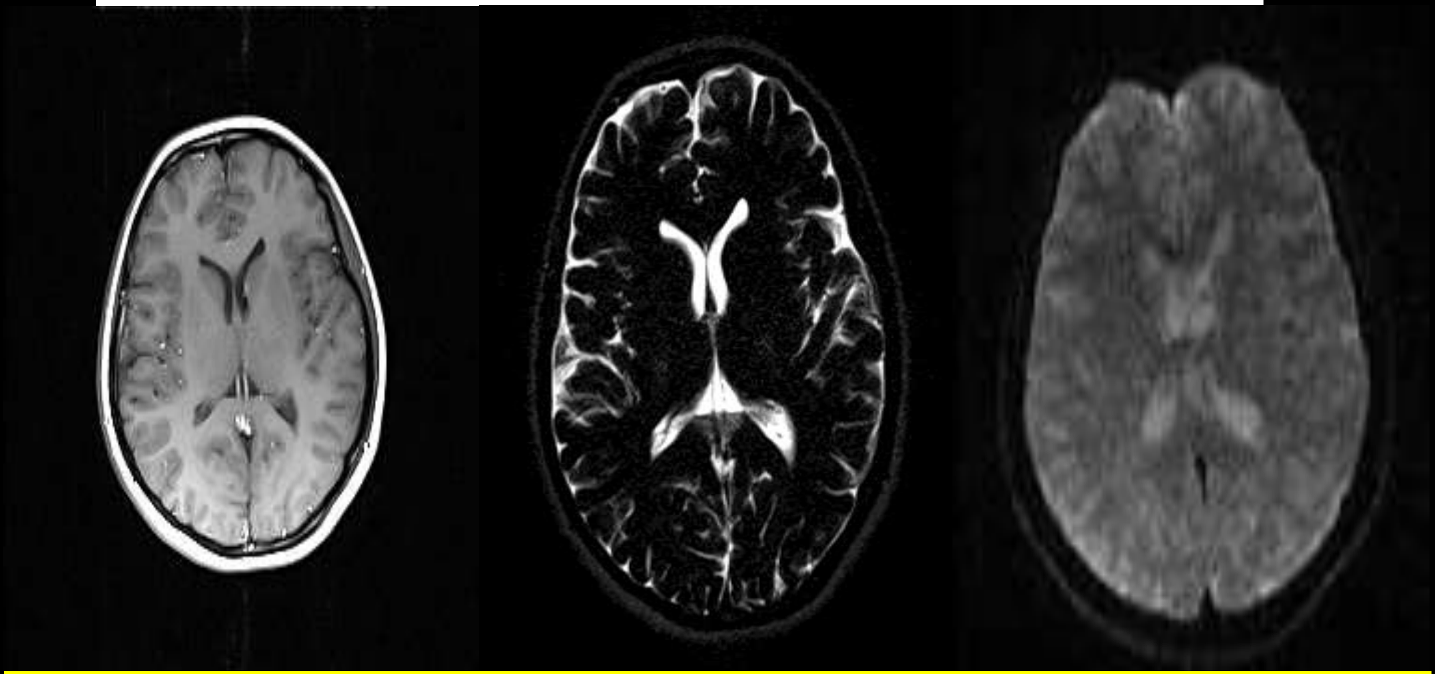
MANIPULATING THE NET MAGNETIZATION

- Magnetization can be manipulated by changing the magnetic field environment (static, gradient, and RF fields)



RELAXATION PROCESSES – BLOCH EQUATIONS & CONTRAST

$$\frac{\partial M_z(t)}{\partial t} = \gamma(\mathbf{M}(t) \times \mathbf{B}(t))_z - \frac{M_z(t) - M_0}{T_1}$$

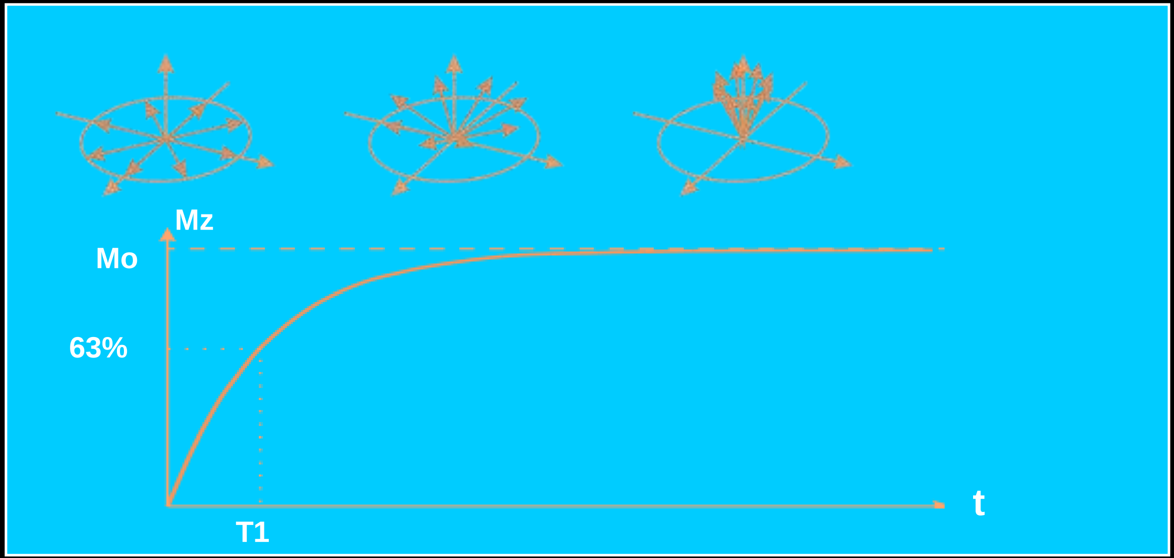


T_1	T_2	DIFFUSION

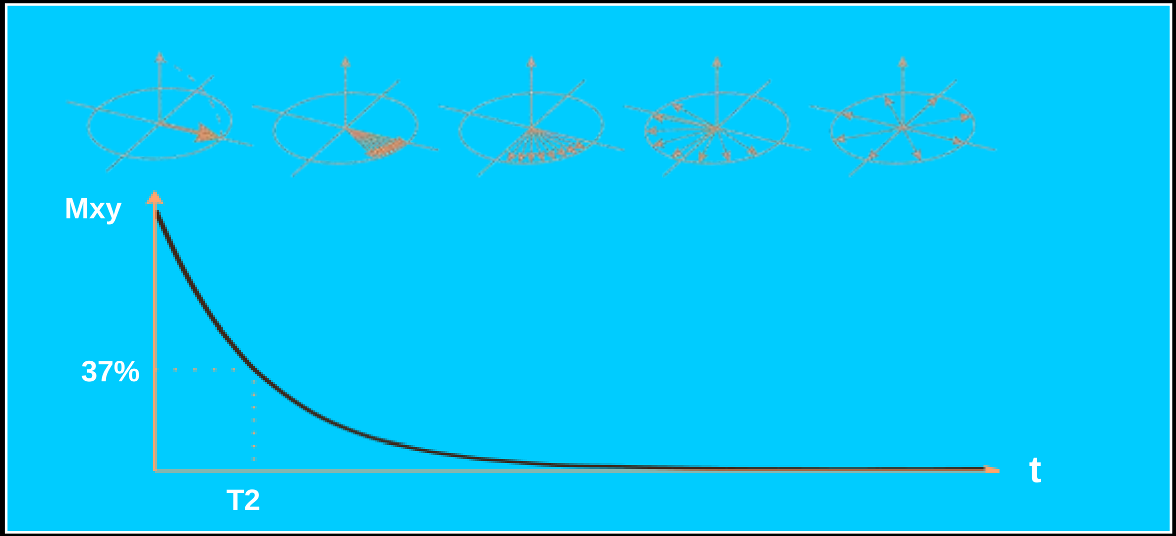
https://www.chemie.uni-hamburg.de/nmr/insensitive/tutorial/en.lproj/vector_model.html
http://www.dayanandasagar.edu/index.php/sharing

Relaxation time

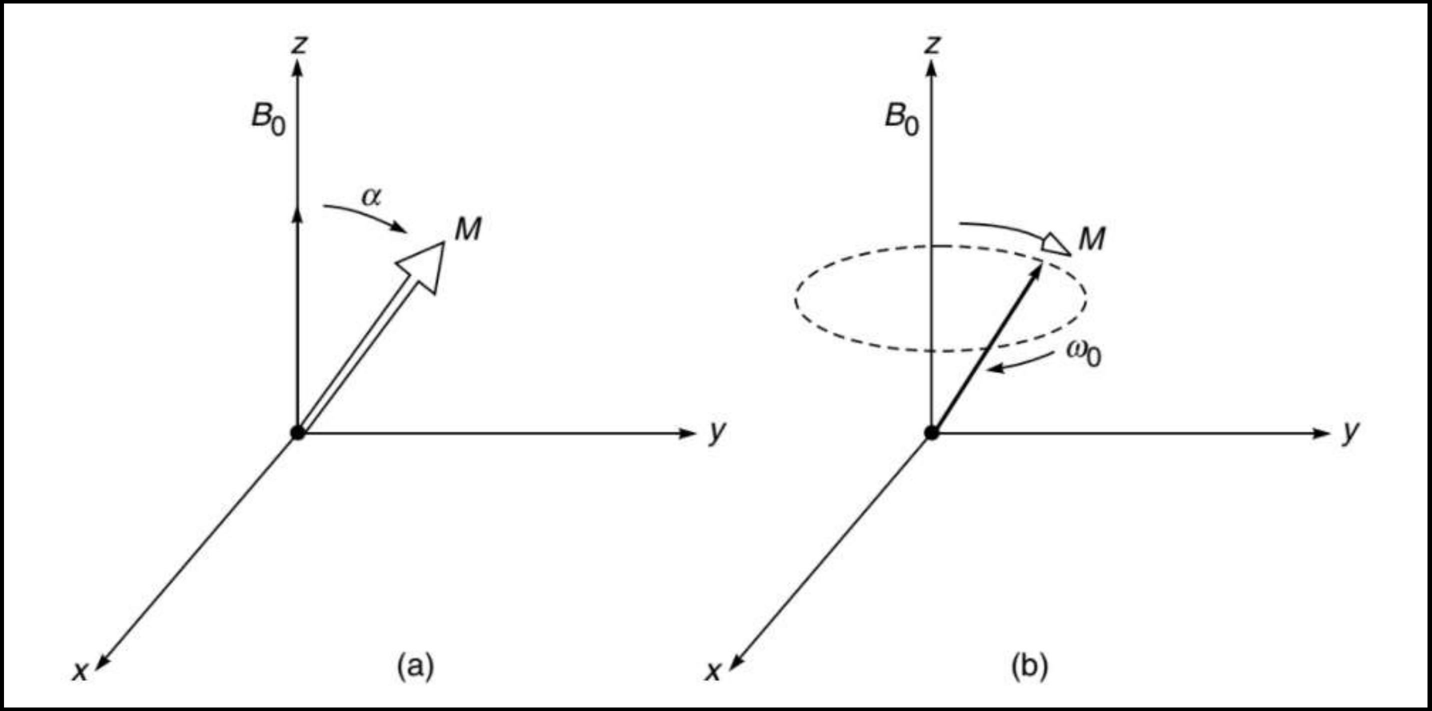
Longitudinal Relaxation Time T_1



Transverse Relaxation Time T_2



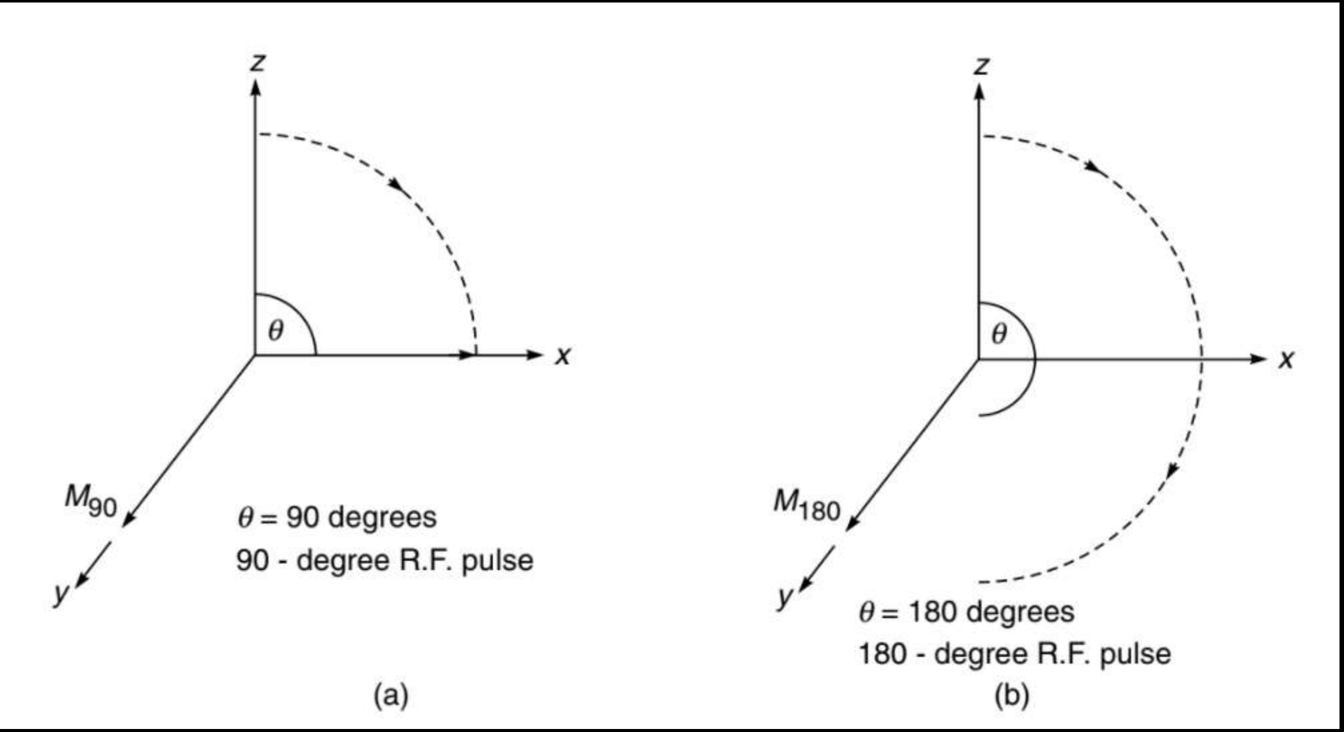
EXCITATION



(a) The magnetic moment is flipped from its equilibrium by the application of another magnetic field

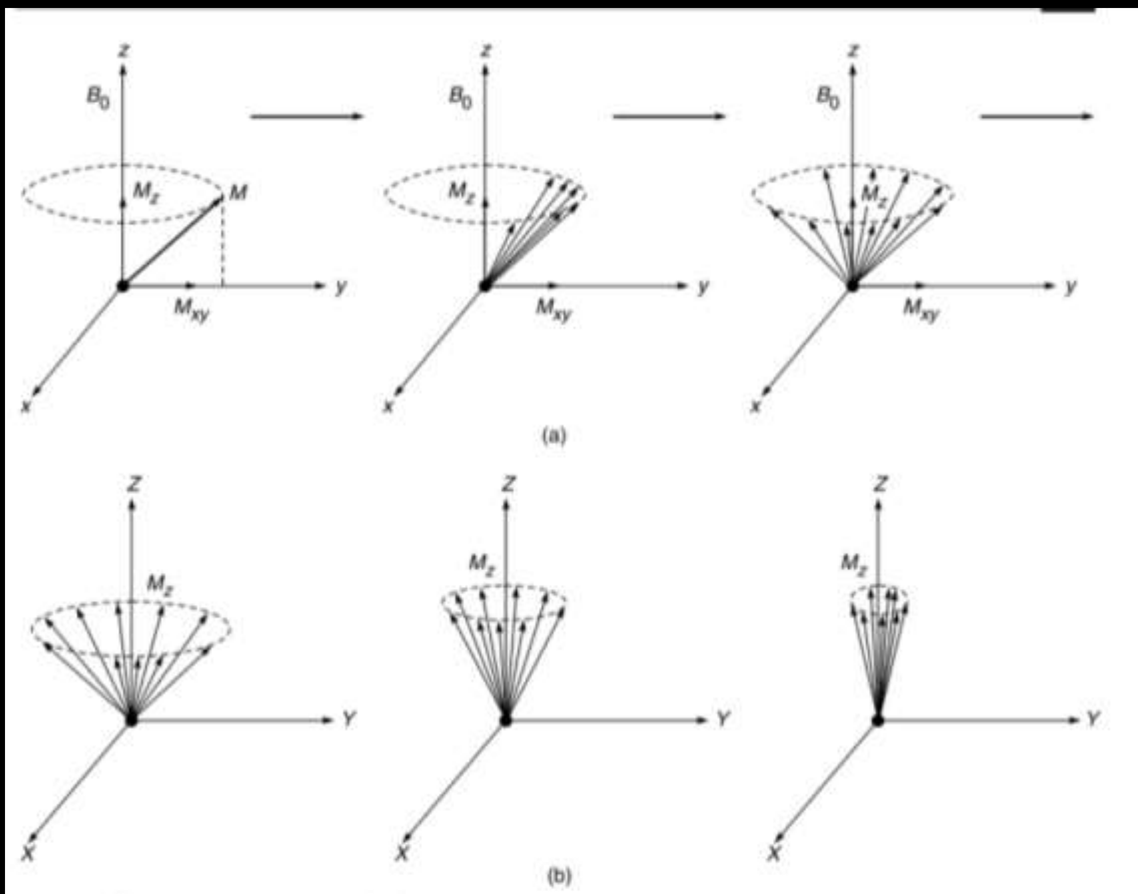
b) It then precesses about the external field direction at a high angular frequency which is proportional to the field strength

EMISSION



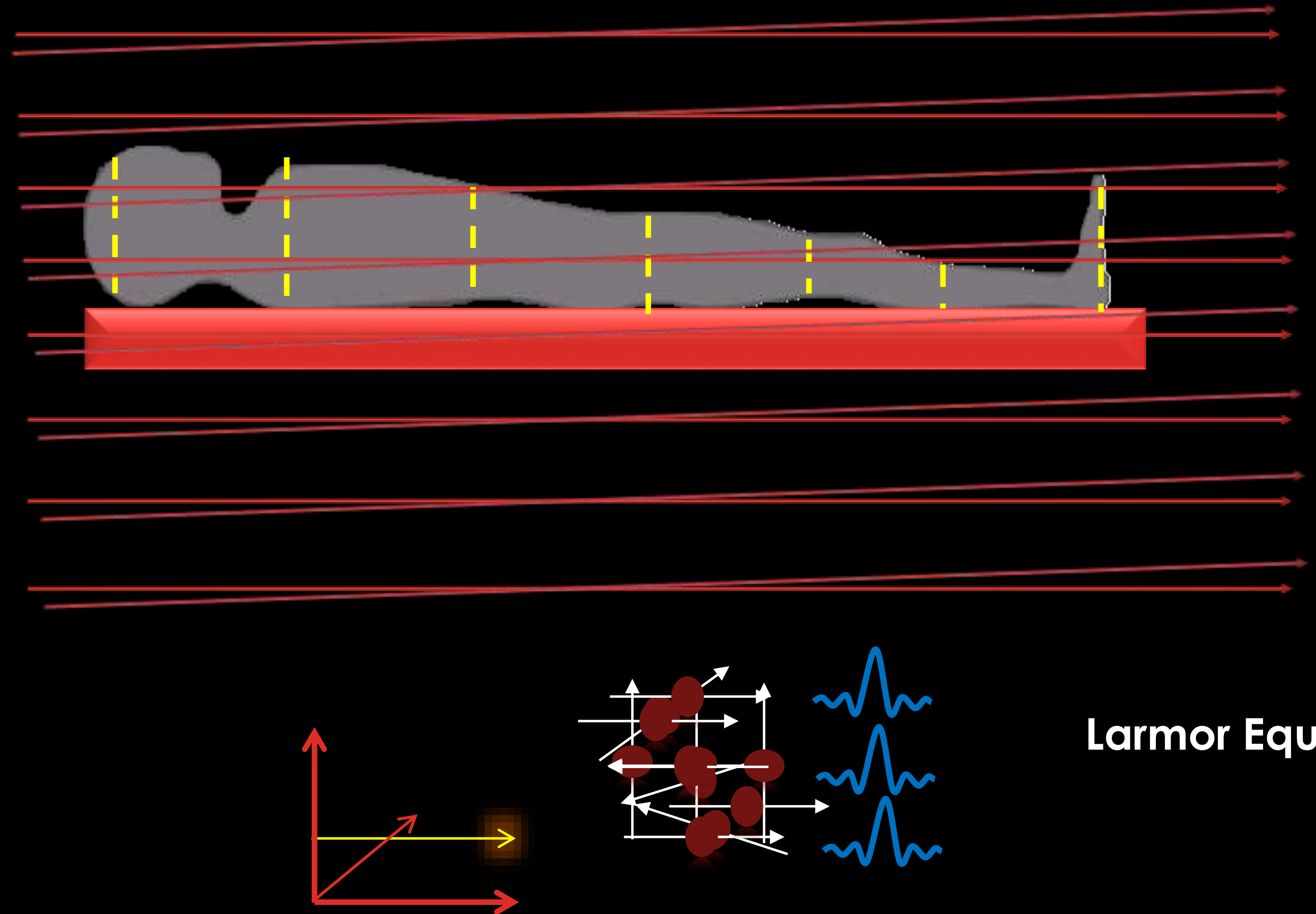
Radio frequency pulse needed to tip the vector of the net magnetic moment M through an angle of (a) 90°, (b) 180° is called 90 and 180 degree pulse respectively

MAGNETIZATION



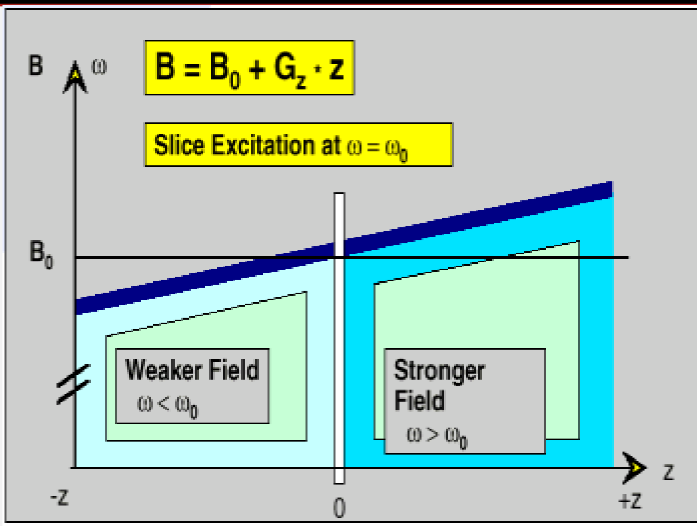
The decay of the magnetization
(a) The signal decays with time constant T2 to give X-Y component
(b) Recovery of the magnetization to its equilibrium position parallel to the
X-axis with a time constant T1

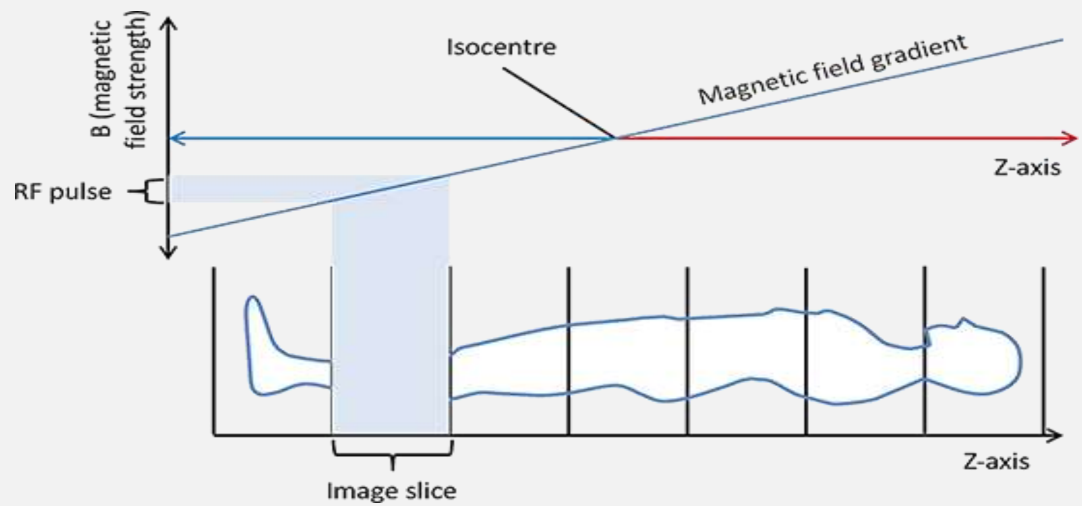
Effect of the B_0, G.r and B_1 fields



Larmor Equation

SLICE SELECTION



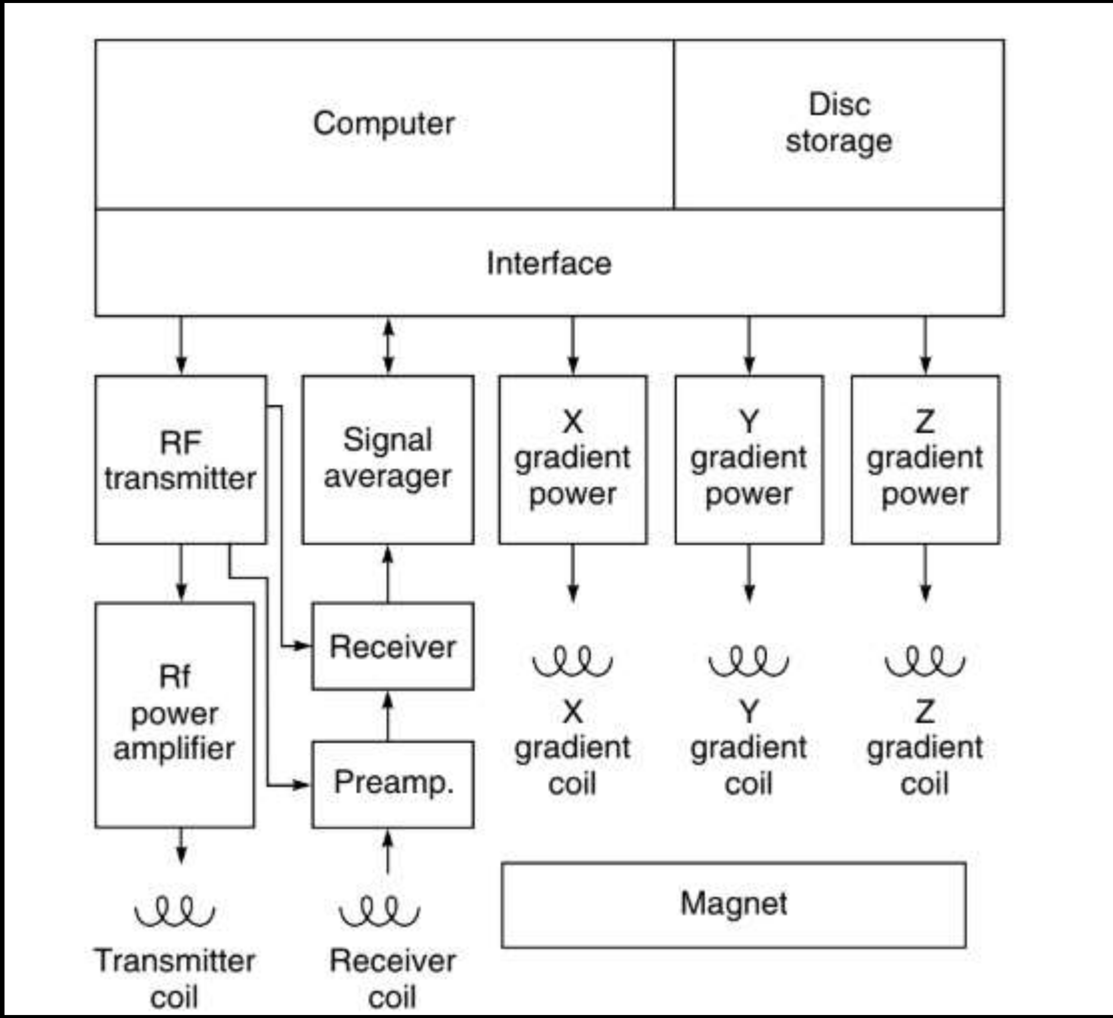


- Slice positions can be changed just by changing the RF pulse frequency
- Slice thickness can be adjusted by bandwidth of the RF

SLICE SELECTION

- The thickness of the slice can be varied by adjusting the bandwidth of the selective pulse and the amplitude of the slice selection gradient:
- For a fixed amplitude gradient, the wider the bandwidth, the greater the number of protons excited and the thicker the slice
- For a fixed bandwidth, the stronger the gradient, the greater the variation of precession frequency in space and the thinner the slice

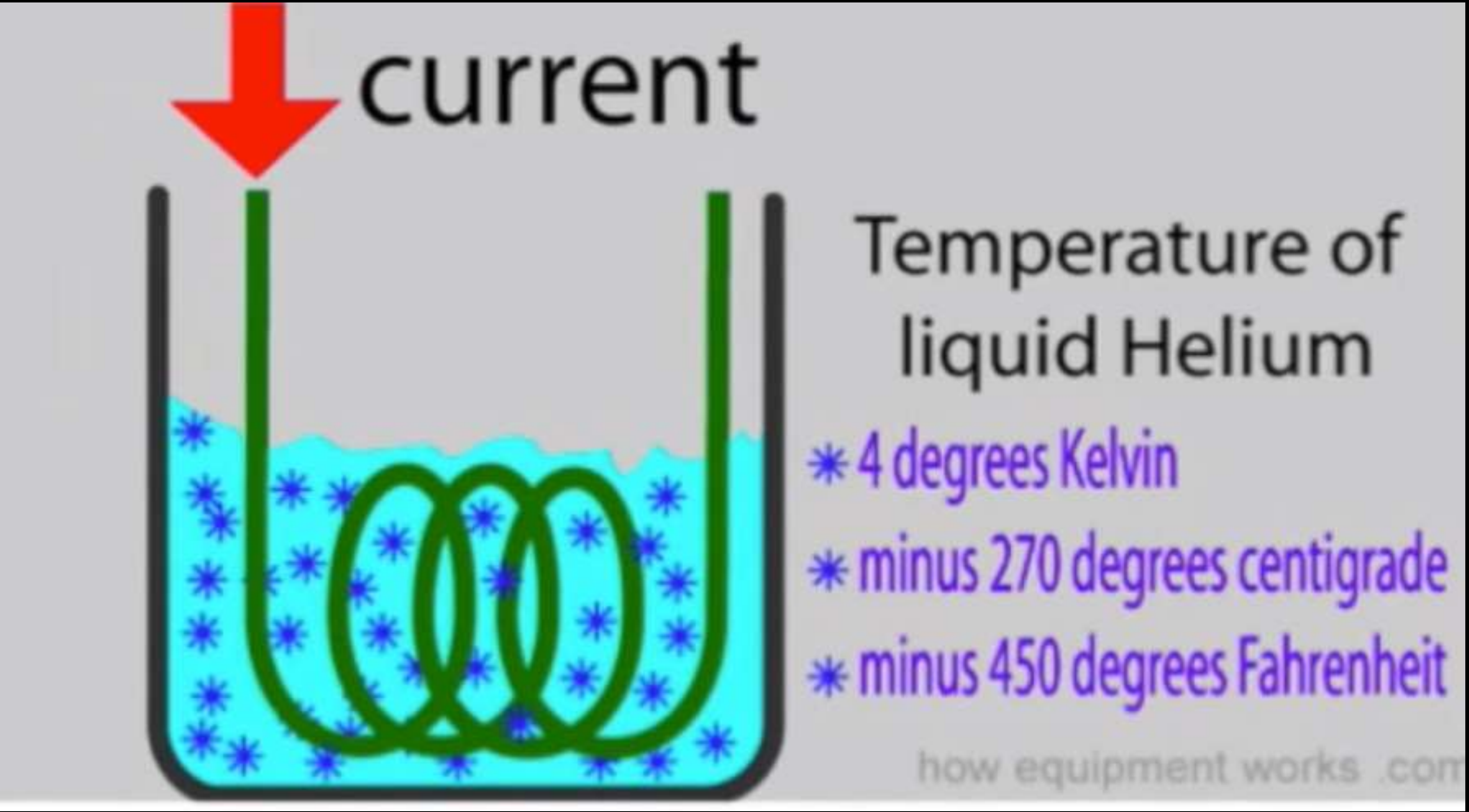
SUBSYSTEMS OF NMR

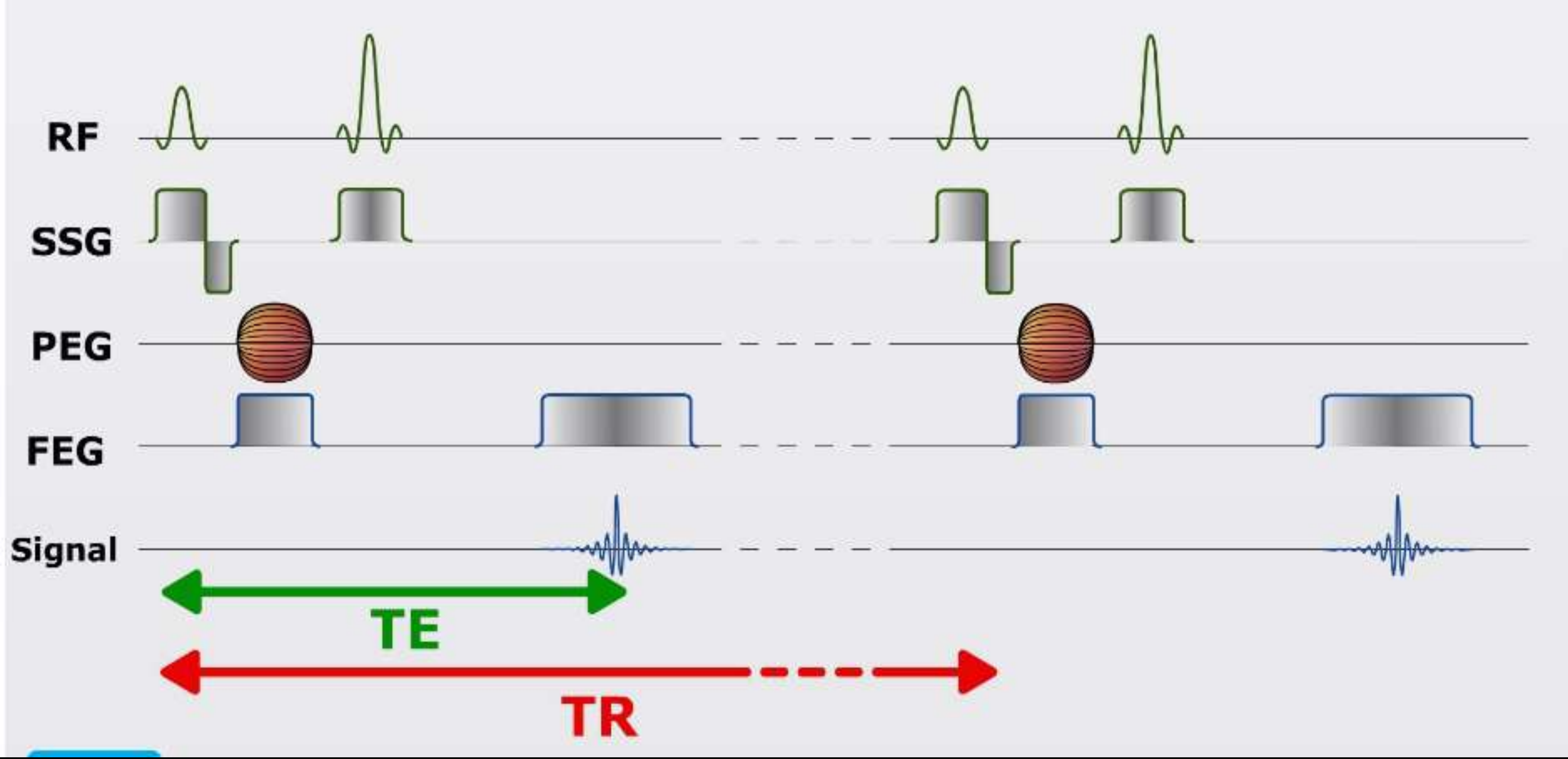


MRI SYSTEM

- A magnet, which provides a strong uniform, steady, magnet field B0
- An RF transmitter, which delivers radio-frequency magnetic field to the sample
- A gradient system, which produces time-varying magnetic fields of controlled spatial non-uniformity
- A detection system, which yields the output signal
- An imager system, including the computer, which reconstructs and displays the images

BORE OF MRI

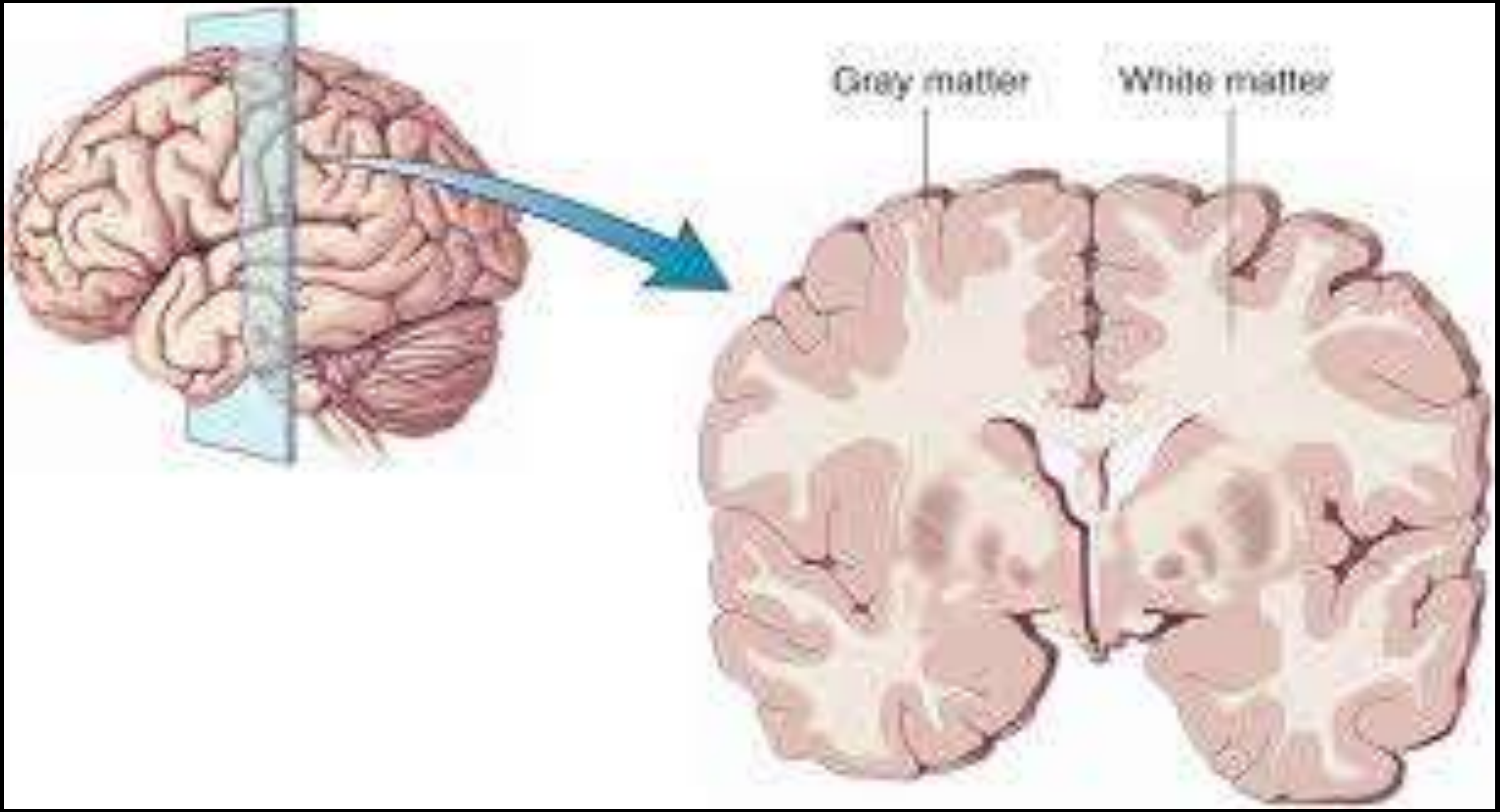

current
Temperature of
liquid Helium
* 4 degrees Kelvin
* minus 270 degrees centigrade
* minus 450 degrees Fahrenheit
how equipment works .com

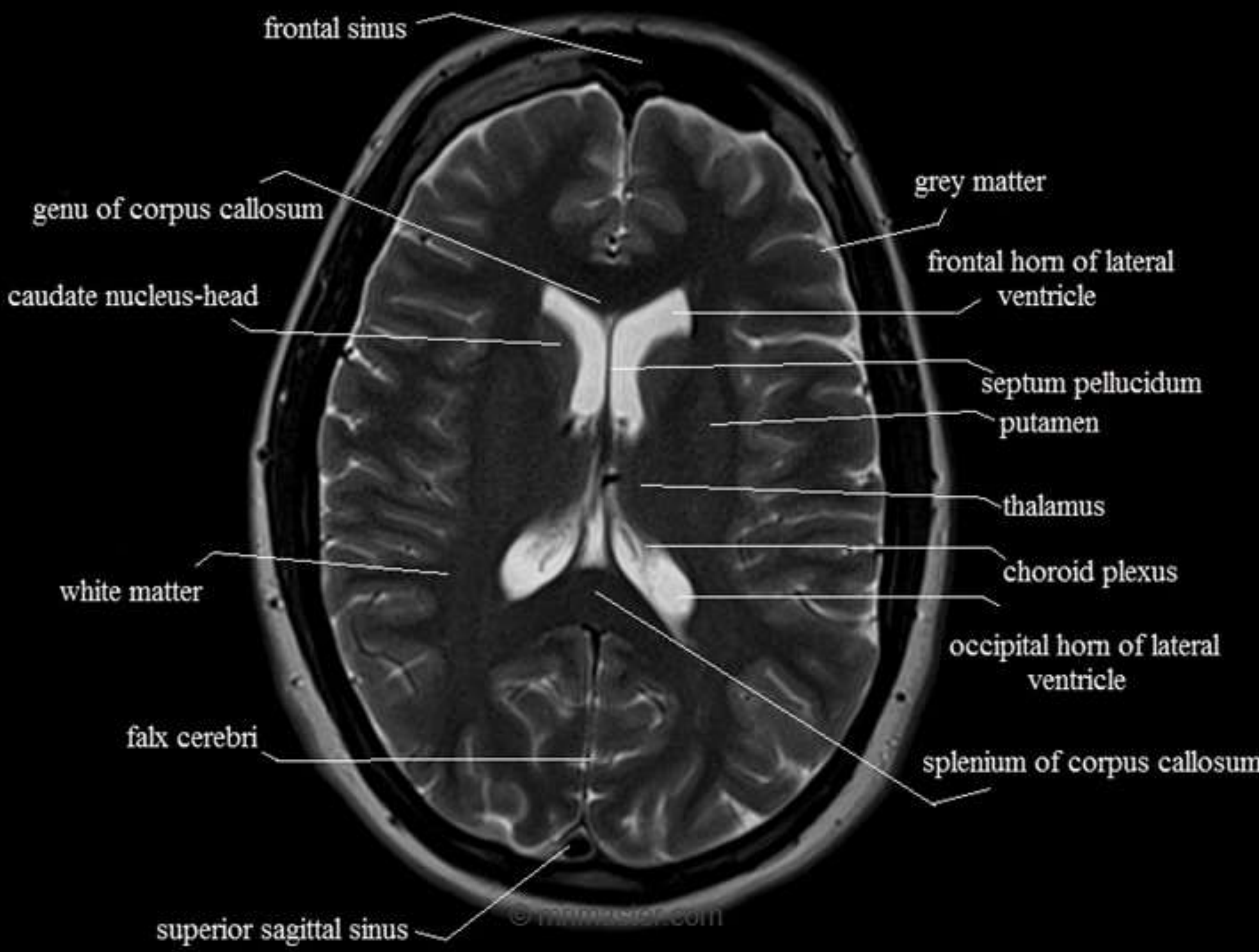


Echo Time (TE) is the time between the 90° RF pulse and MR signal sampling, corresponding to maximum of echo. (The 180° RF pulse is applied at time TE/2- In Spin Echo imaging).

Repetition Time (TR) is the time between 2 excitations pulses (time between two 90° RF pulses).

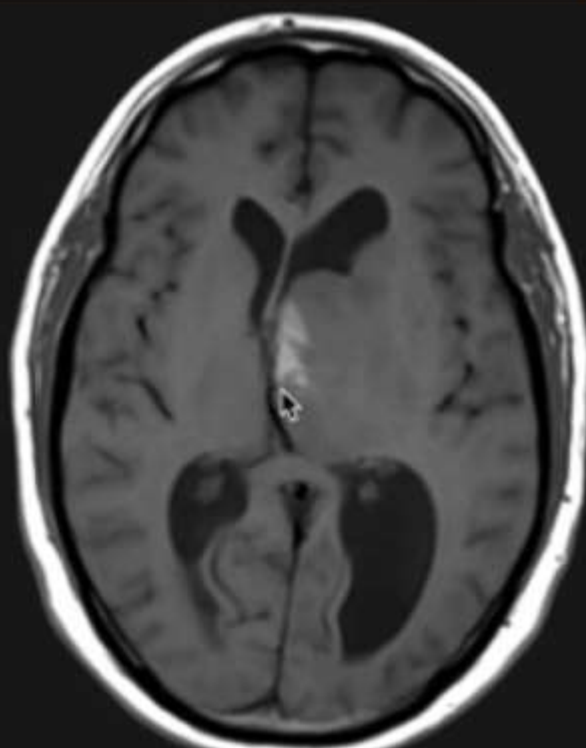
STRUCTURE OF BRAIN




frontal sinus
genu of corpus callosum
caudate nucleus-head
white matter
falx cerebri
superior sagittal sinus
grey matter
frontal horn of lateral ventricle
septum pellucidum
putamen
thalamus
choroid plexus
occipital horn of lateral ventricle
splenium of corpus callosum

T1 weighted

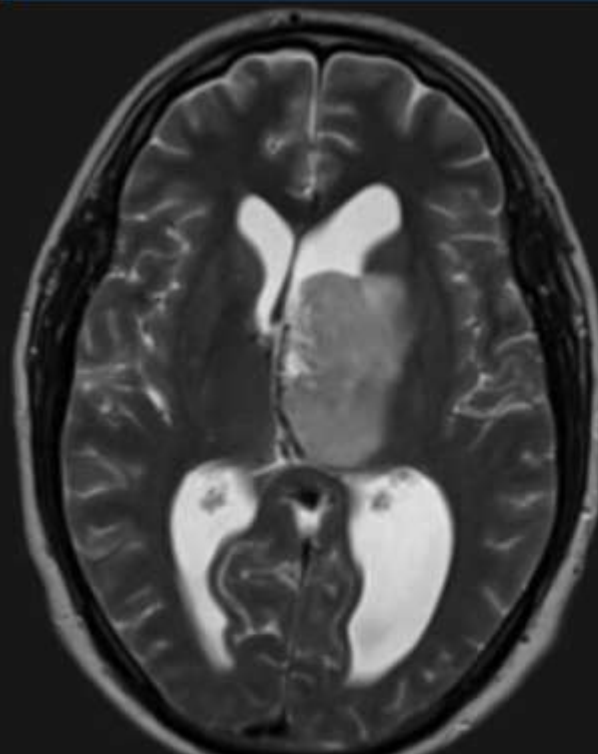
TR short
TE short



Dr Frank Gaillard rID 14249

T2 weighted

TR long
TE long



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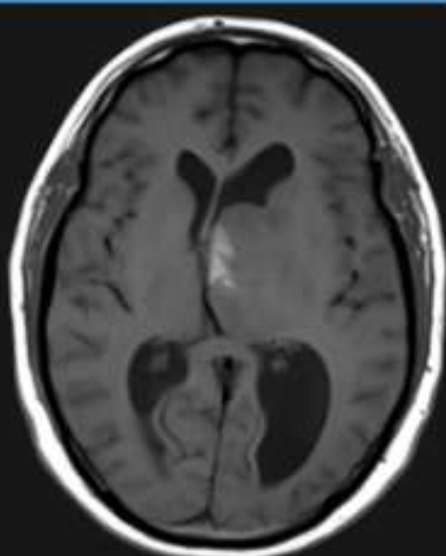
Sequences

T1 weighted	Proton density	T2 weighted
TR short TE short	TR long TE short	TR long TE long



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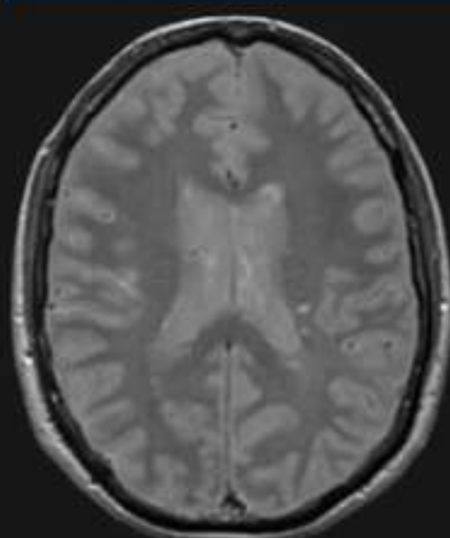
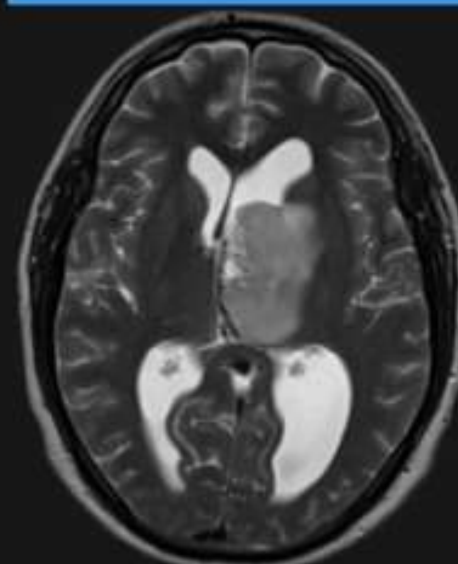


Dr Frank Gaillard (unpublished)



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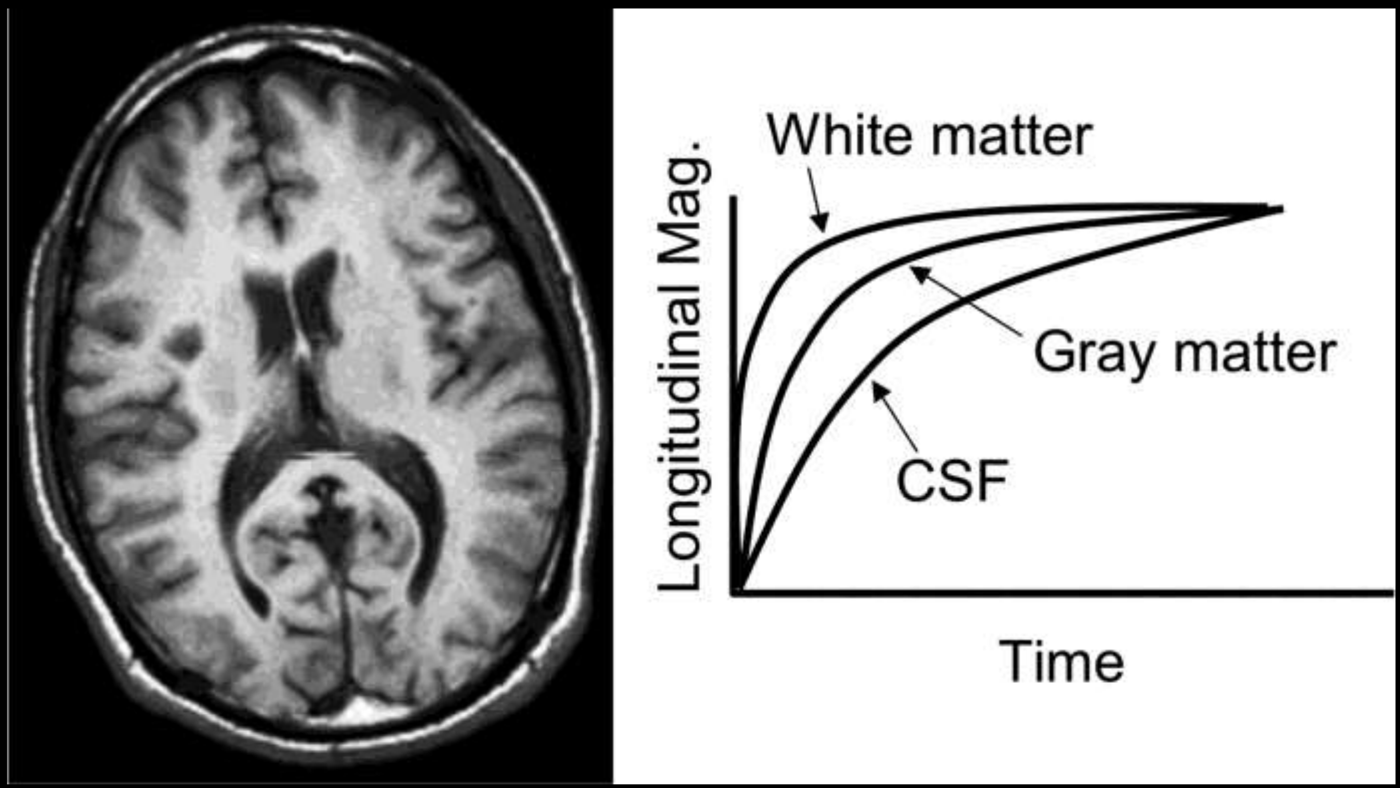
MR SEQUENCES

- Choosing the right sequence parameters (TR and TE) will produce images weighted in T1, T2 or proton density.

- Historically, spin echo was the first sequence to be used. It has been a benchmark for all subsequent developments, namely in terms of contrast.

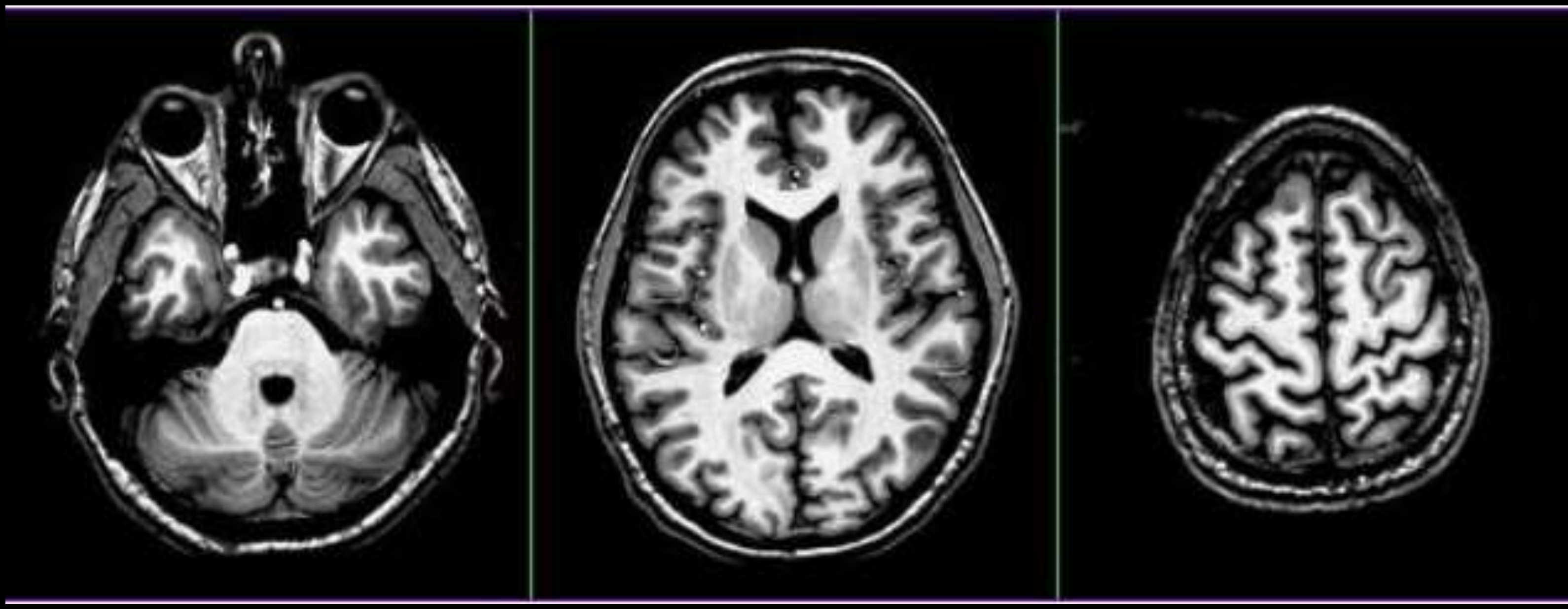
- While spin echo sequences can be used in clinical practice to obtain good quality anatomical T1-weighted images, faster types of sequence are preferred to obtain T2-weighted images.

T1-weighted contrast

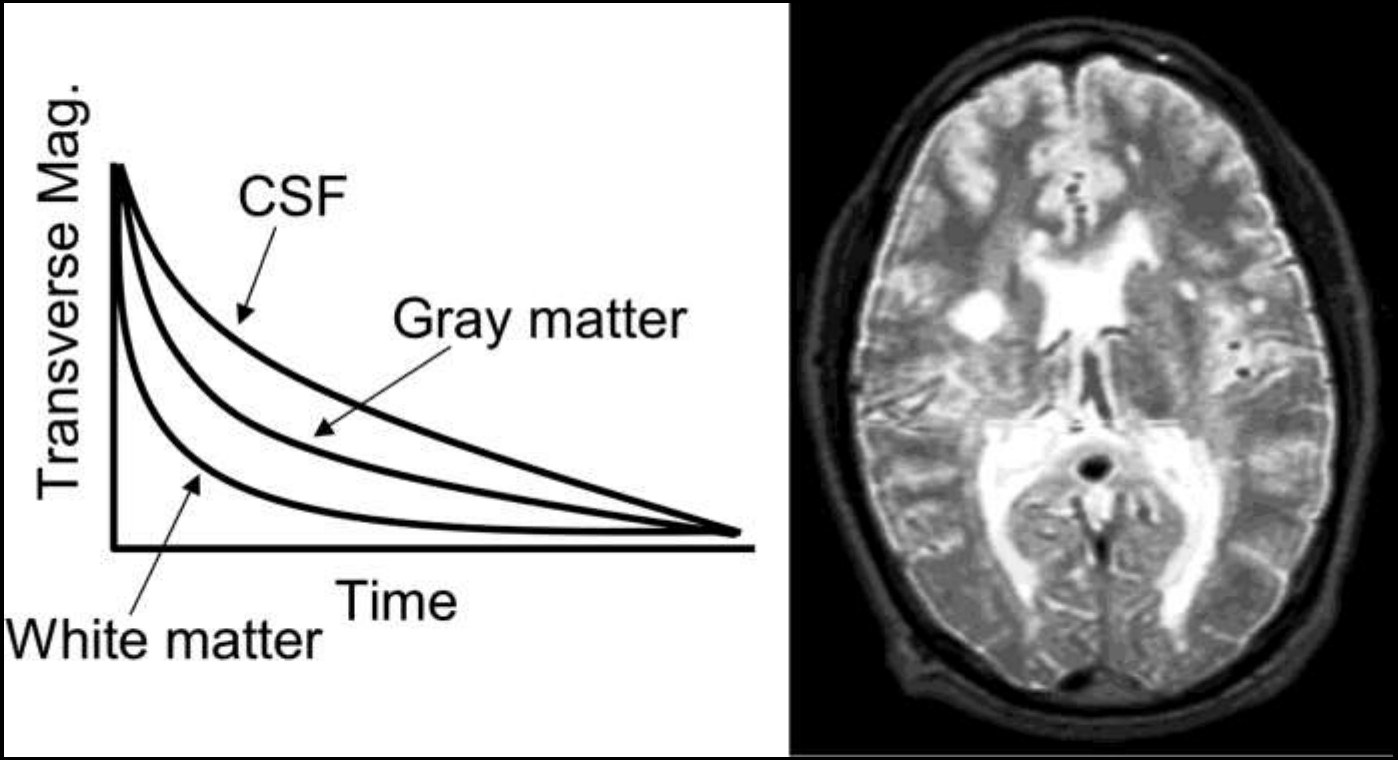


Pooley, R. A. Radiographics 2005;25:1087-1099

T_1 WEIGHTED IMAGING

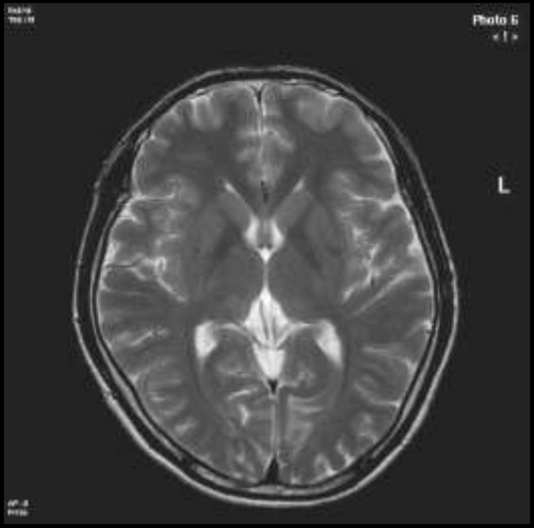


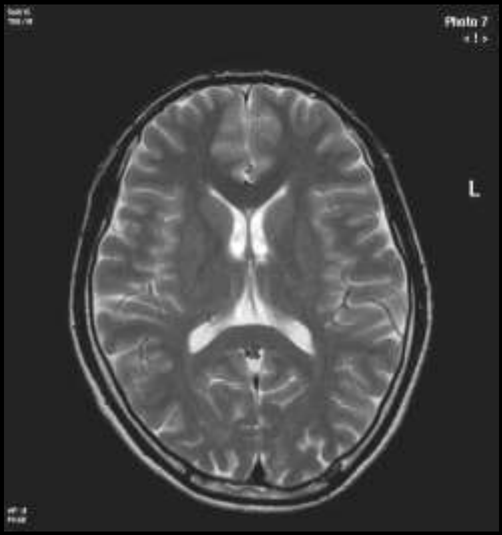
T_2 weighted contrast



Pooley, R. A. Radiographics 2005;25:1087-1099

T_2 WEIGHTED IMAGING

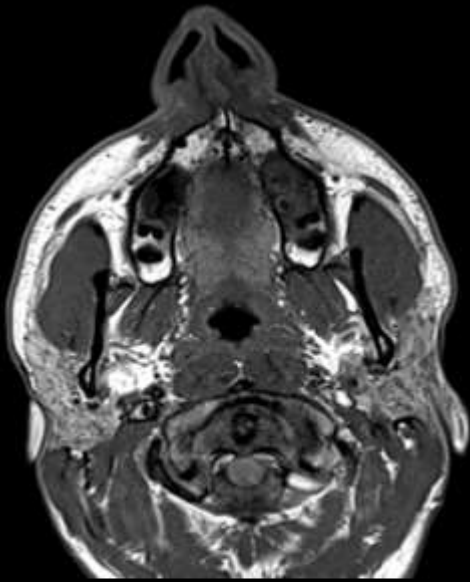


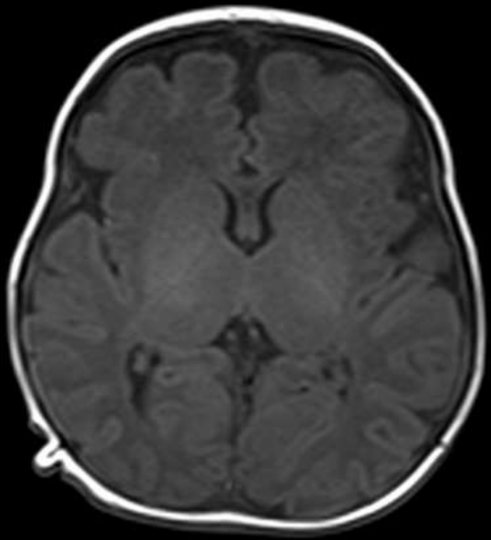


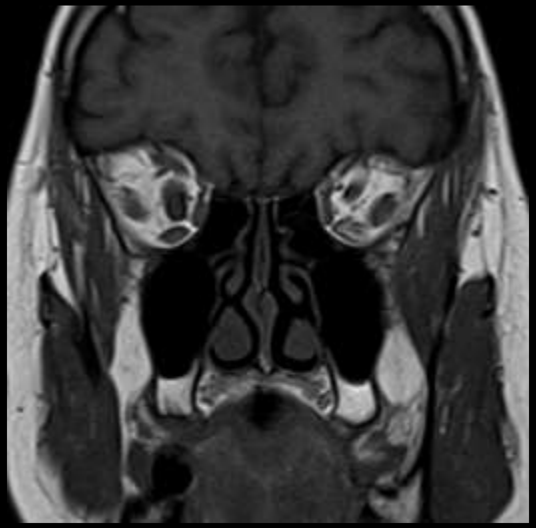
Proton density imaging

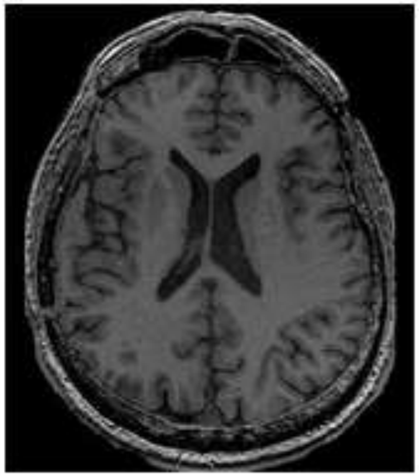
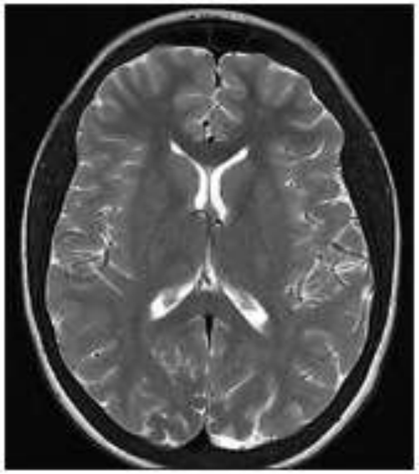
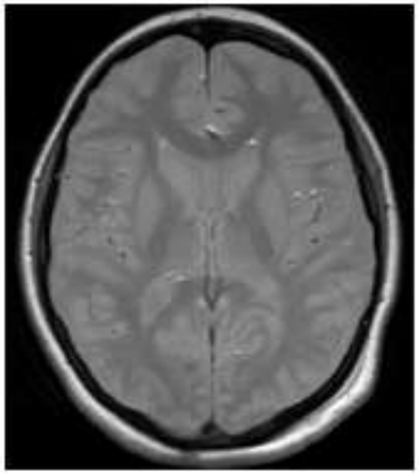
- Proton density (PD) does not display the magnetic characteristics of the hydrogen nuclei but the number of nuclei in the area being imaged
- To get a PD weighted image we have to minimize the contribution of both T1 and T2 contrast
- T1 minimized with a long TR: large signal and small T1 contrast
- T2 minimized with a short TE: large signal and small T2 contrast

Proton density imaging







	T1	T2	PD
Image			
Water signal	Water has a long T1. T1-WI uses a short TR so the signal from water is still low, therfore, water appears **dark**	T2-WI uses a long TE so the signal from water is high, therefore, water appears **bright**	A long TR results in a high water signal, but a short TE means that this is less than the signal of a T2 scan. The signal of water is in the middle
Fat signal	Fat has a short T1, so even though the TR is short the signal is still high and fat appears **bright**	Fat has a short T2 so at a long TE the signal is **less bright** and it will be darker than water	A long TR results in a high fat signal and short TE means this signal is higher than on a T2-WI: fat appears **bright**
TR	Short. 300-600 ms	Long. 2000 ms	Long. 1000-3000 ms
TE	Short. 10-30 ms	Long. 90-140 ms	Short. 15 ms

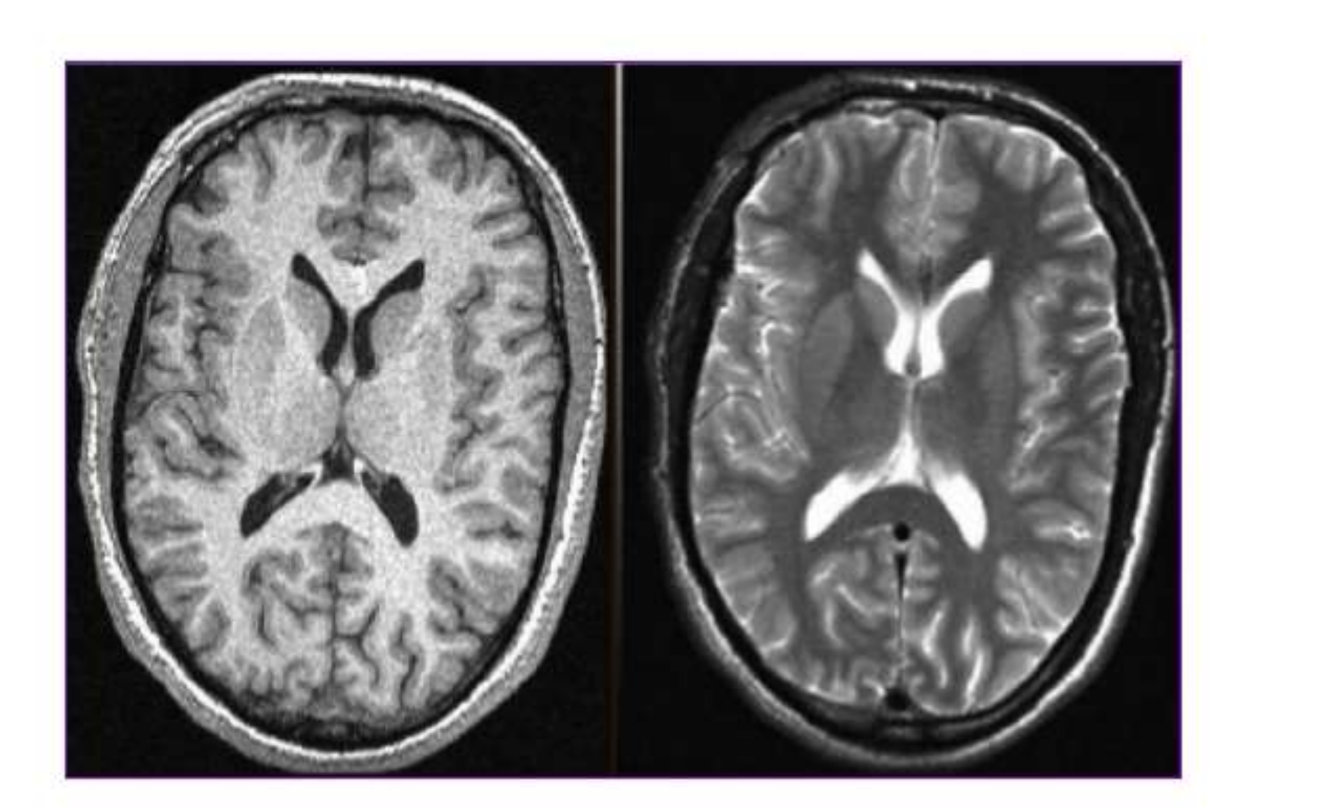
https://www.radiologycafe.com/

TISSUE APPEARANCE

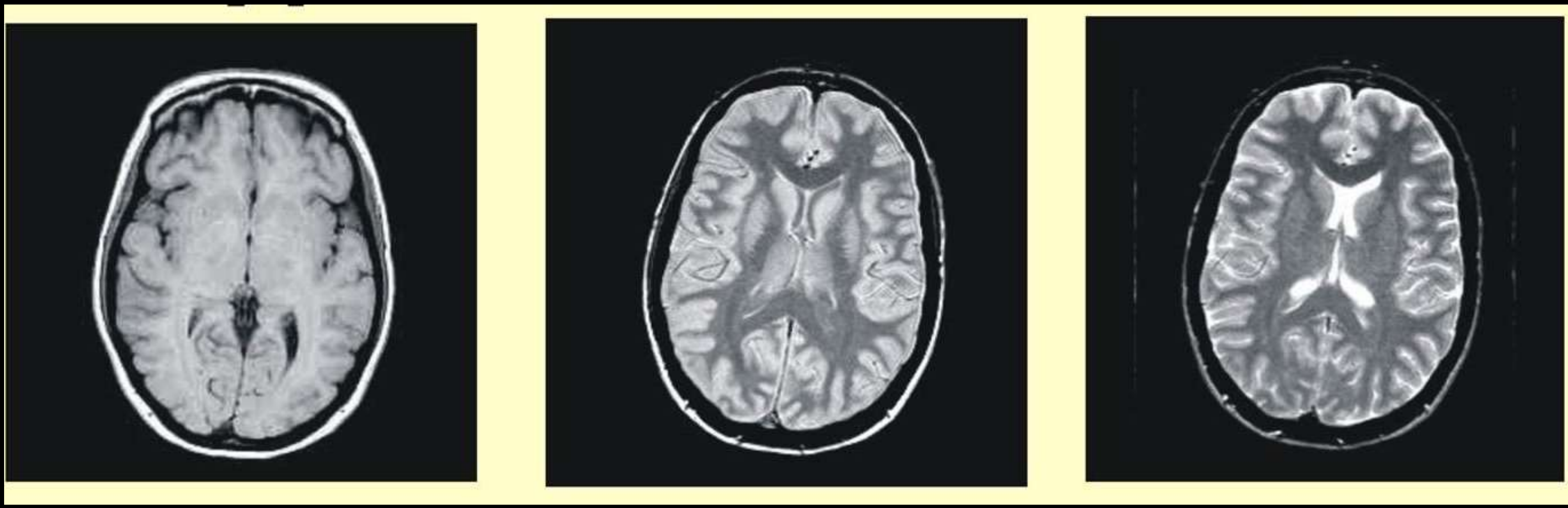
WT	FAT	H2O	MUSC	LIG	BONE
T1	B	D	I	D	D
Proton Density	I	I	I	D	D
T2	I	B	I	D	D

IMAGING						
CT SCAN	CSF	Edema	White Matter	Gray Matter	Blood	Bone
MRI T1	CSF	Edema	Gray Matter	White Matter	Cartilage	Fat
MRI T2	Cartilage	Fat	White Matter	Gray Matter	Edema	CSF
MRI PD	Bone	Cartilage	edema	White Matter	Gray Matter	Fat

IDENTIFY THE PROTOCOL!!



IDENTIFY THE PROTOCOL!!

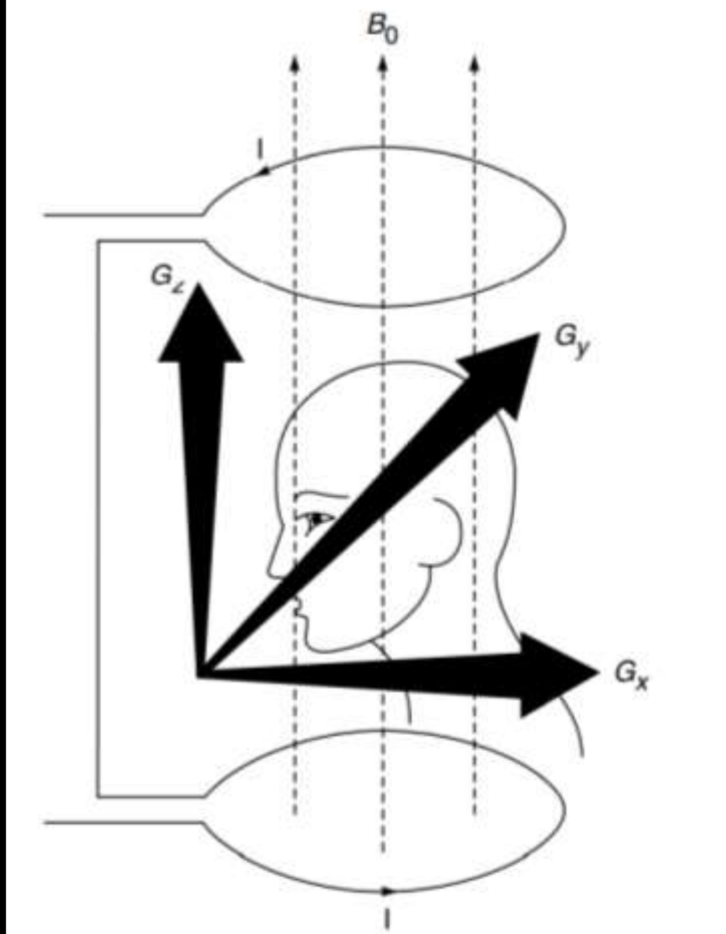


SPATIAL INFORMATION

- For imaging it is necessary to add spatial data to the signal to allocate a position to the different signals
- To do so, we start by selecting the slice plane, within which the horizontal and vertical directions will then be defined
- The term “encoding ” is used, as the spatial data obtained are not the classic co-ordinates (x, y, z), but are observed through a specific spatial filter
- The RF signals received are then processed to reverse the filter effect and reconstruct the image
- Decoding of spatial information, included in the NMR signal as modifications of frequency and phase, is performed by an inverse Fourier Transform

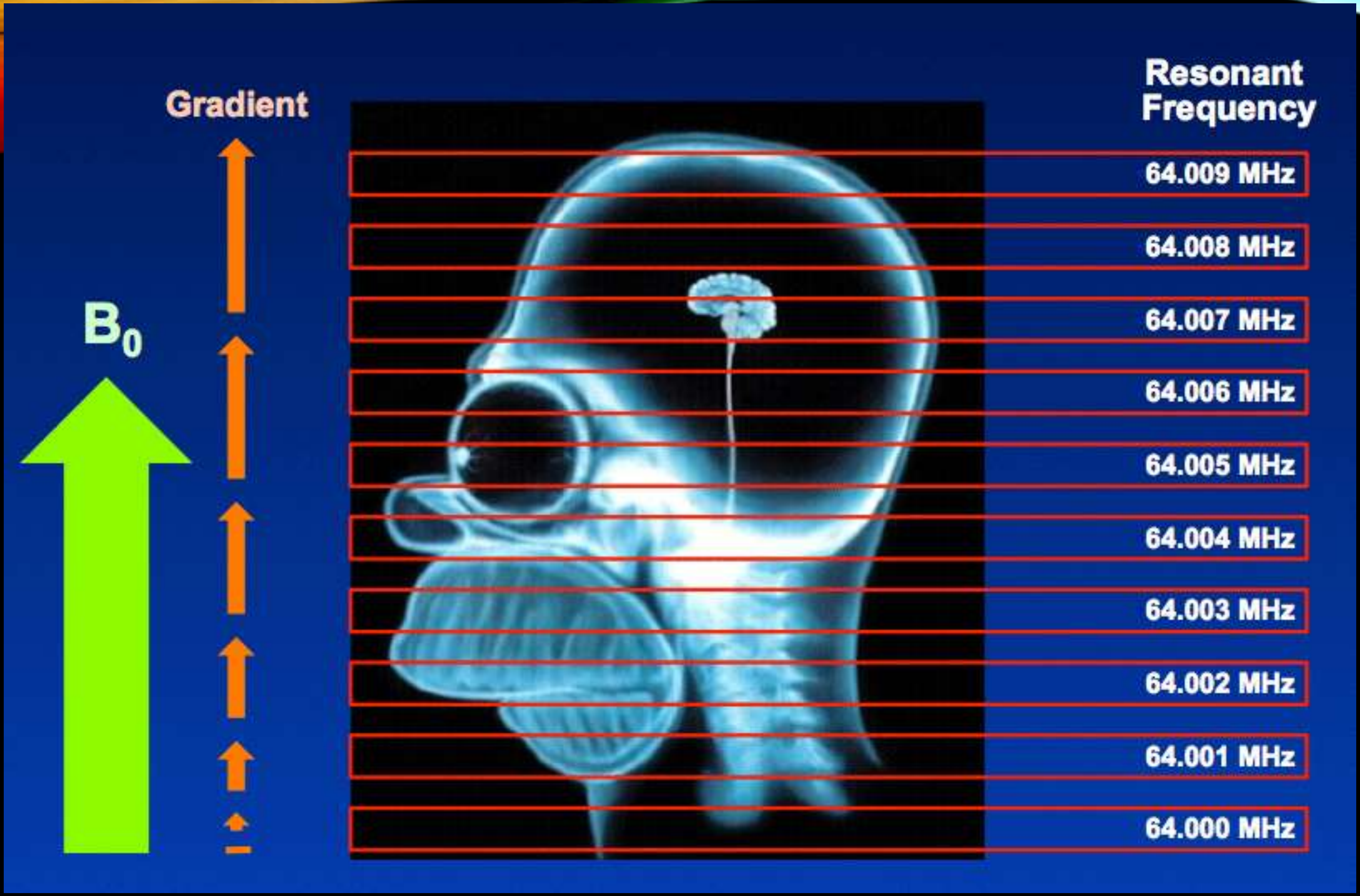
SPATIAL ENCODING

- Spatial encoding relies on successively applying magnetic field gradients
- First of all, a slice selection gradient (G_{SS}) is used to select the anatomical volume of interest
- Within this volume, the position of each point will be encoded vertically and horizontally by applying a phase encoding gradient (GPE), and a frequency-encoding gradient (GFE)



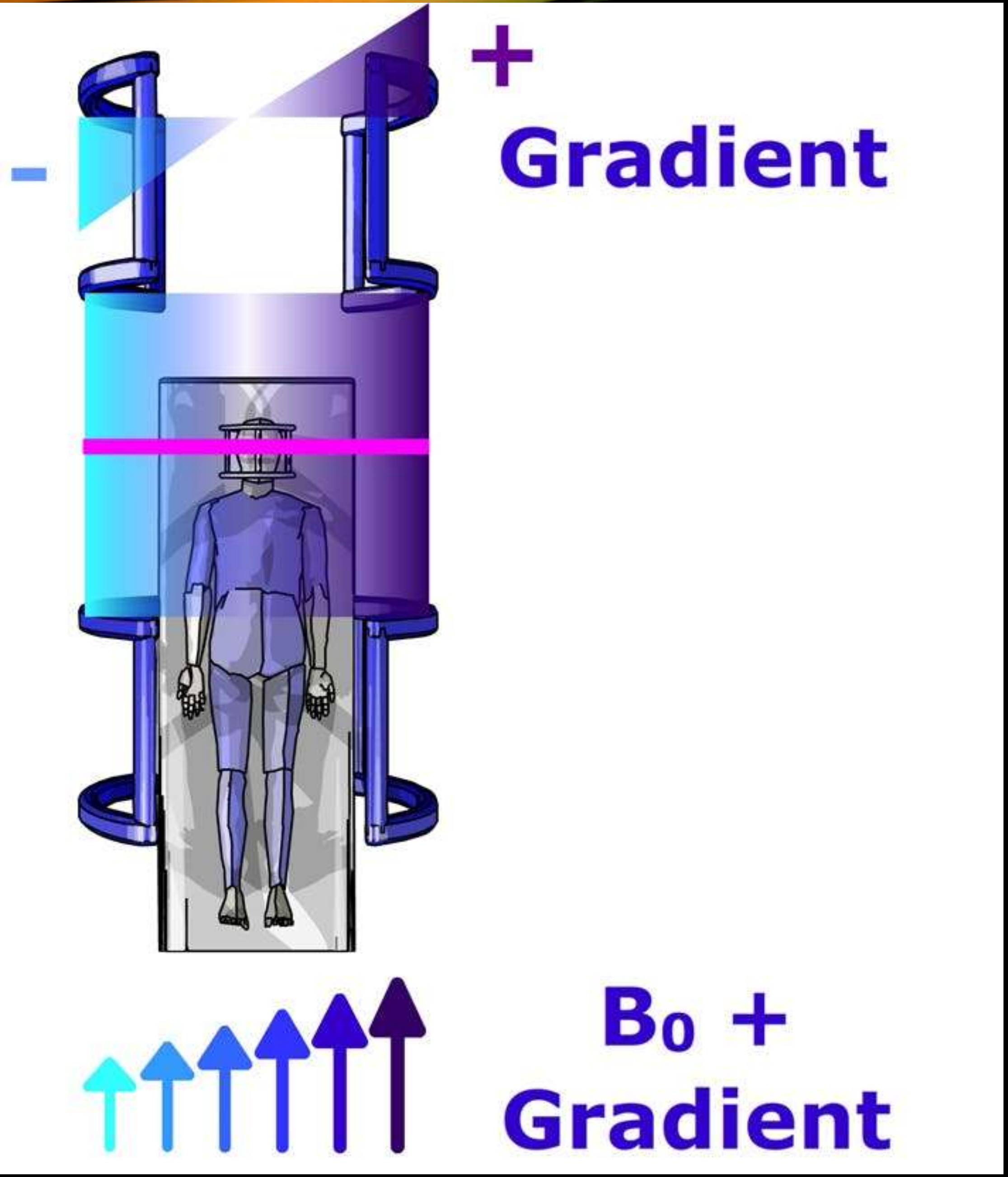
➤ Fig. 22.22 *Arrangement of the field gradients*

SLICE SELECTION

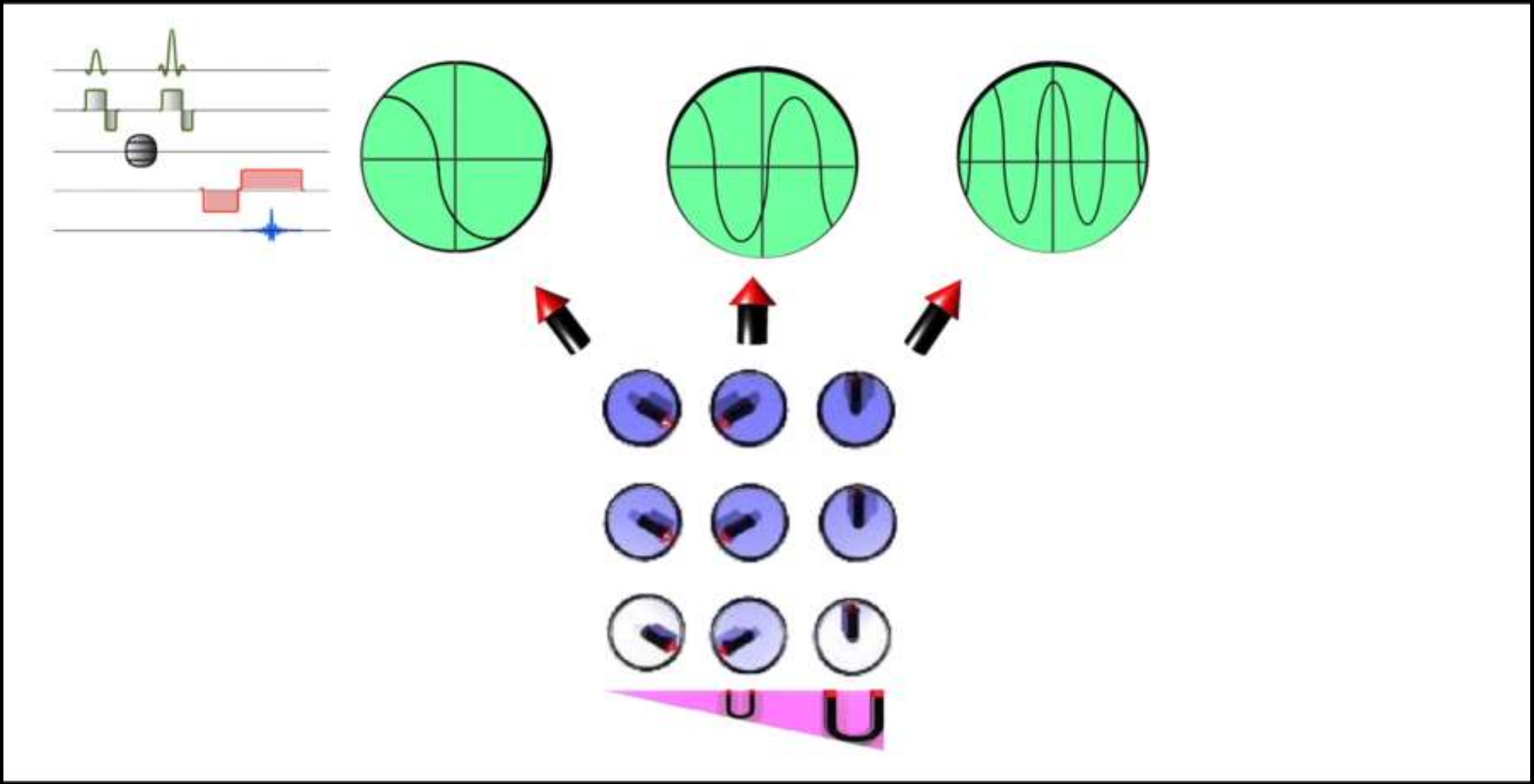


MAGNETIC FIELD GRADIENT APPLIED IN THE Z-DIRECTION CAUSES THE RESONANT FREQUENCIES TO VARY BY A FEW THOUSAND HZ FROM SLICE TO SLICE.

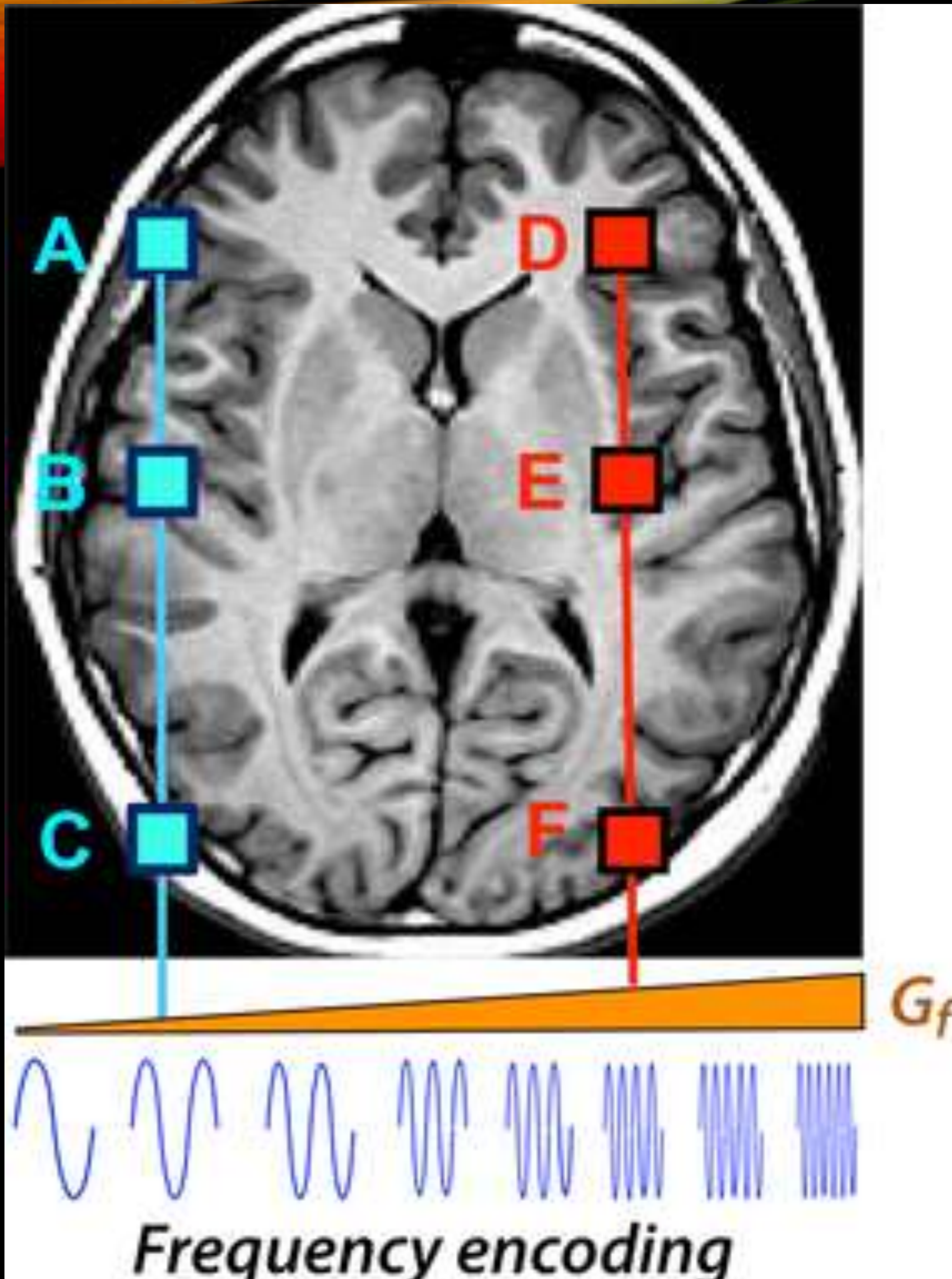
FREQUENCY ENCODING



FREQUENCY ENCODING



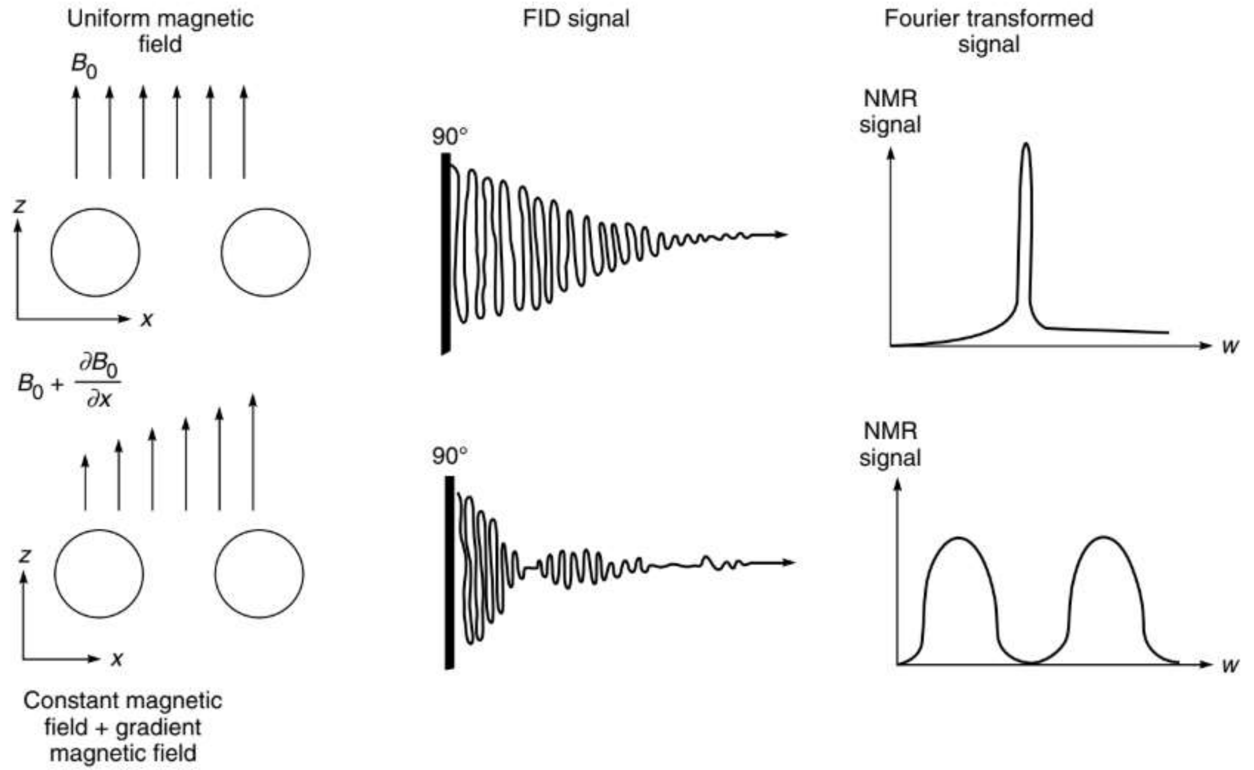
FREQUENCY ENCODING



If the main (static) magnetic field is Bo, then the effective field $B(x)$ at any point (x) along the horizontal axis is given by:
$B(x) = Bo + xGf$

From the ***Larmor equation*** ($f = \gamma B$), we can easily show that the resonant frequency $f(x)$ also varies linearly with position (x) along the frequency-encoding axis:
$f(x) = \gamma B(x) = \gamma Bo + \gamma x \bullet Gf = fo + fg(x)$
where fo is the main field Larmor frequency and $fg(x)$ is the frequency offset based on position along the gradient.

FOURIER TRANSFORM OF FID



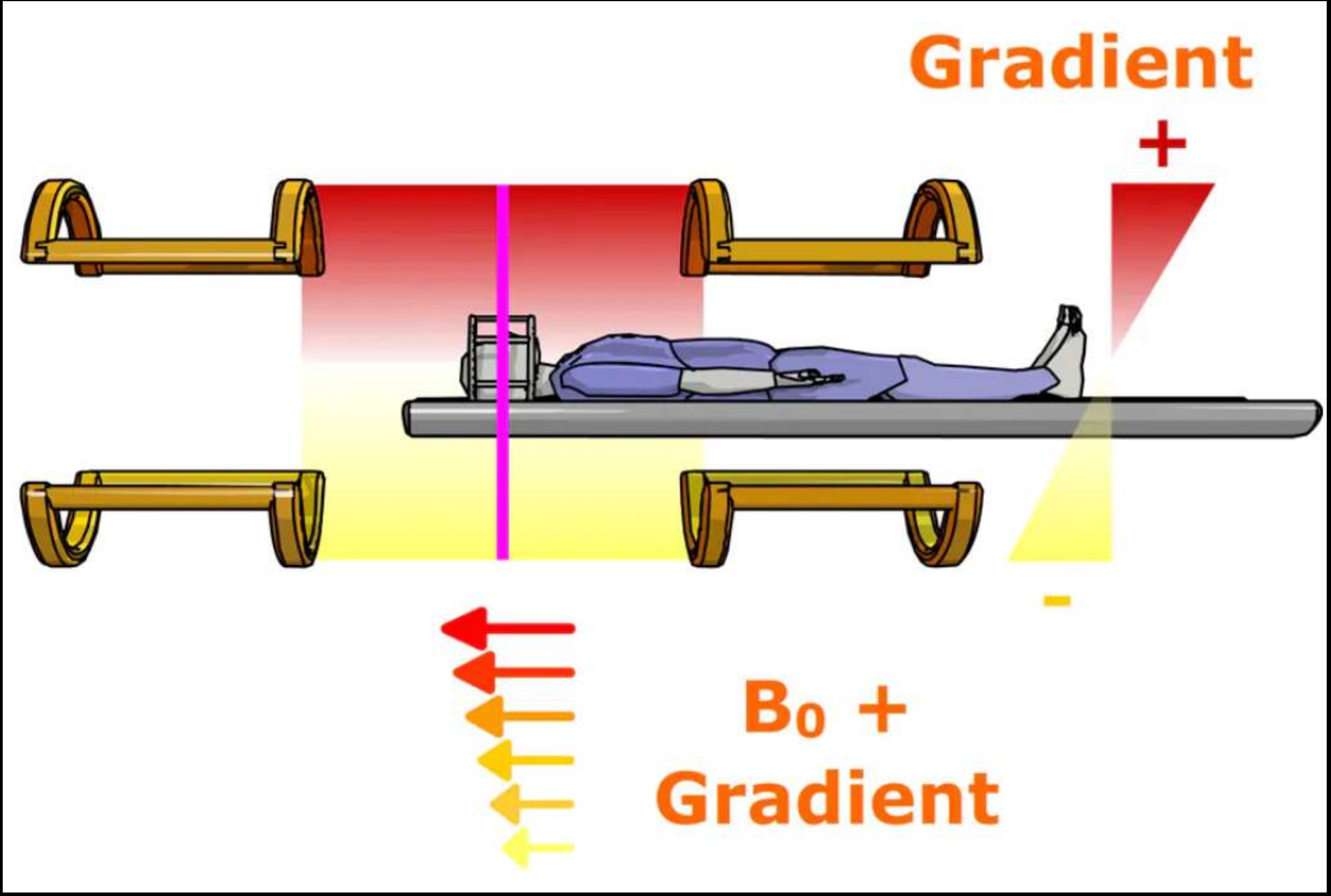
➤ **Fig.** 22.8 *Fourier transformation of the FID signal*

The Fourier transformation $f(\omega)$ of a function of time $f(t)$ is given by

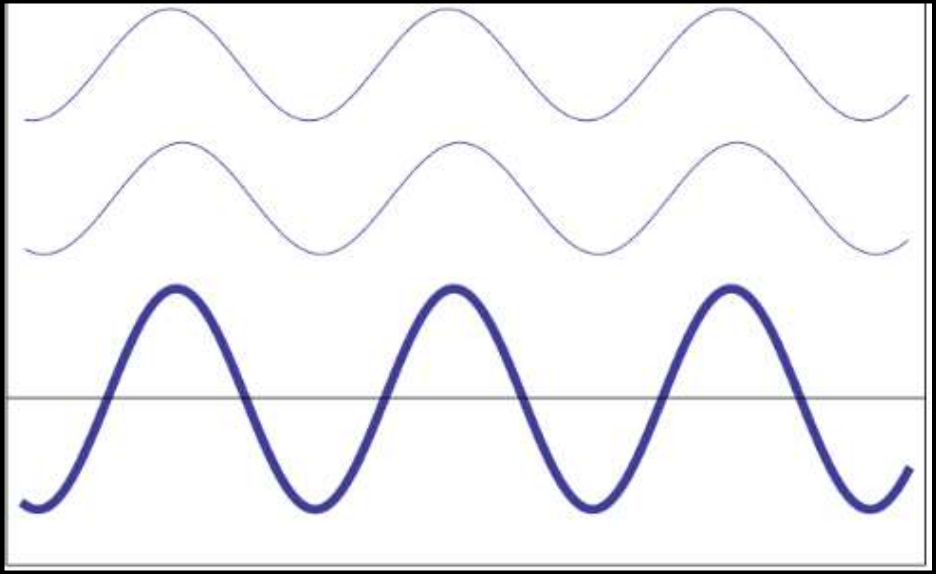
$$f(\omega) = \int_{-\infty}^{\infty} f(t)e^{-i\omega t}\, dt$$

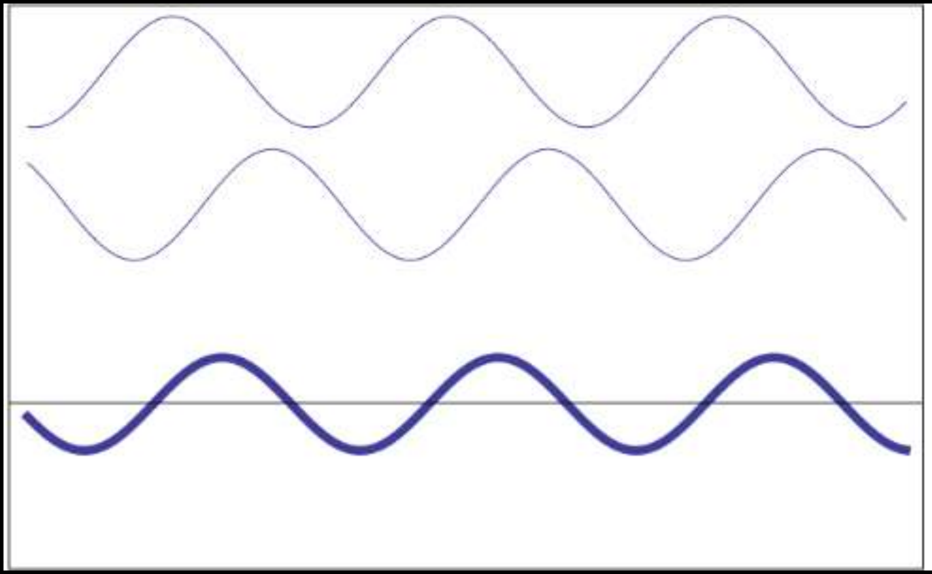
where ω is the angular frequency ($\omega = 2\pi v$) and $i = \sqrt{-1}$

PHASE ENCODING



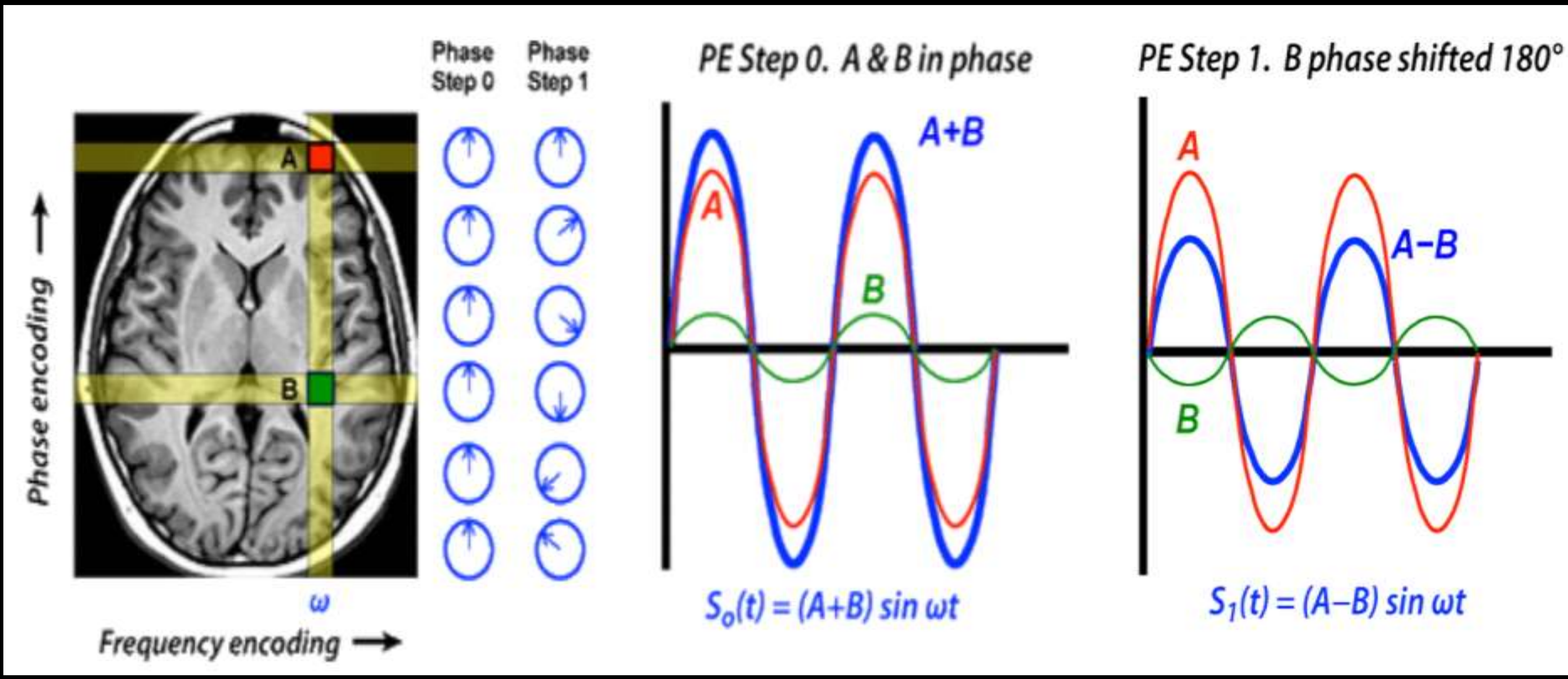
PHASE ENCODING



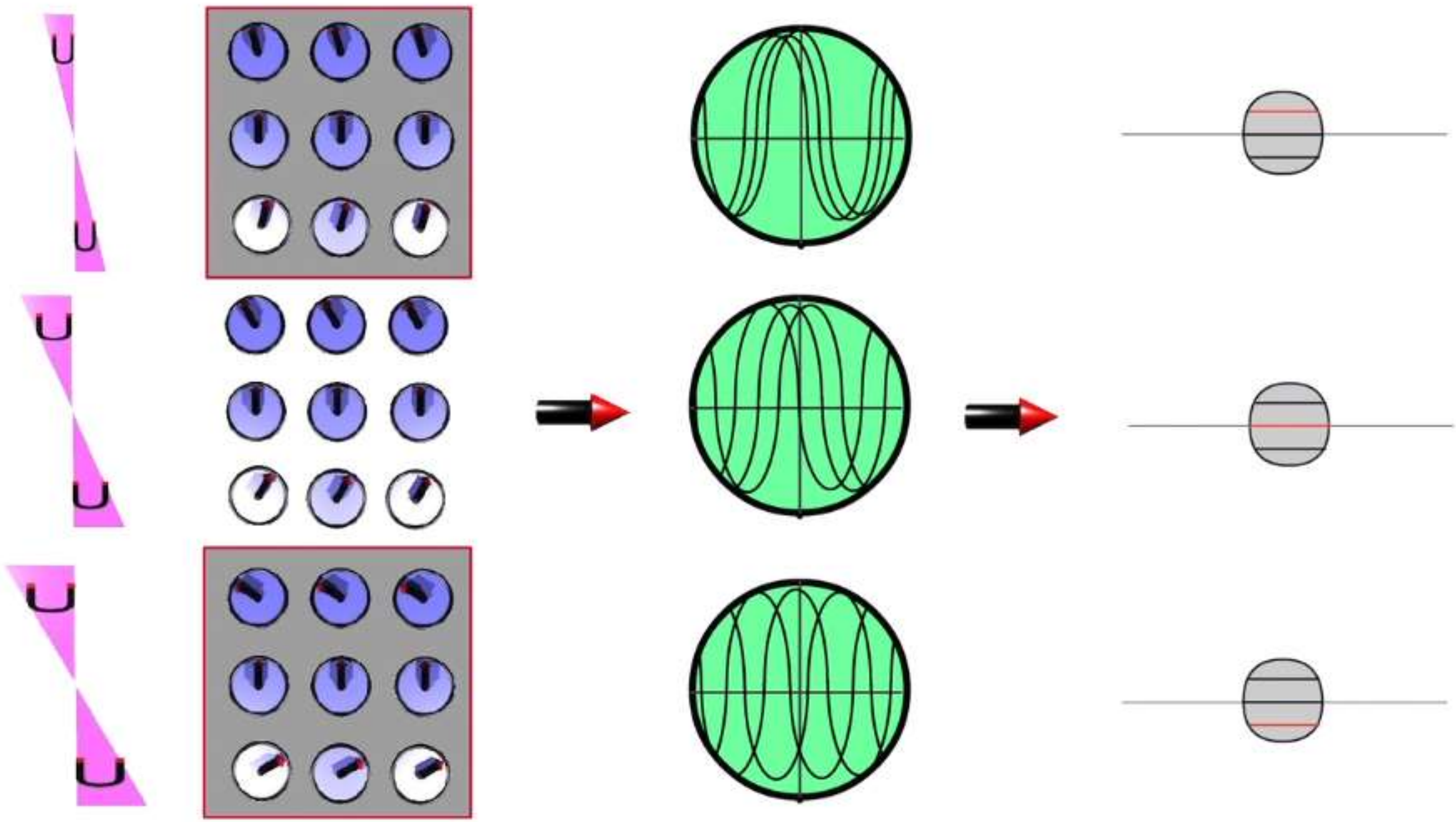


when two sine waves (A and B) with the same frequency but different phases are added together, the result is another sine wave with the same frequency but a different phase. When the sine waves are close together in phase they constructively interfere, and when out of phase they destructively interfere

PHASE ENCODING



PHASE ENCODING

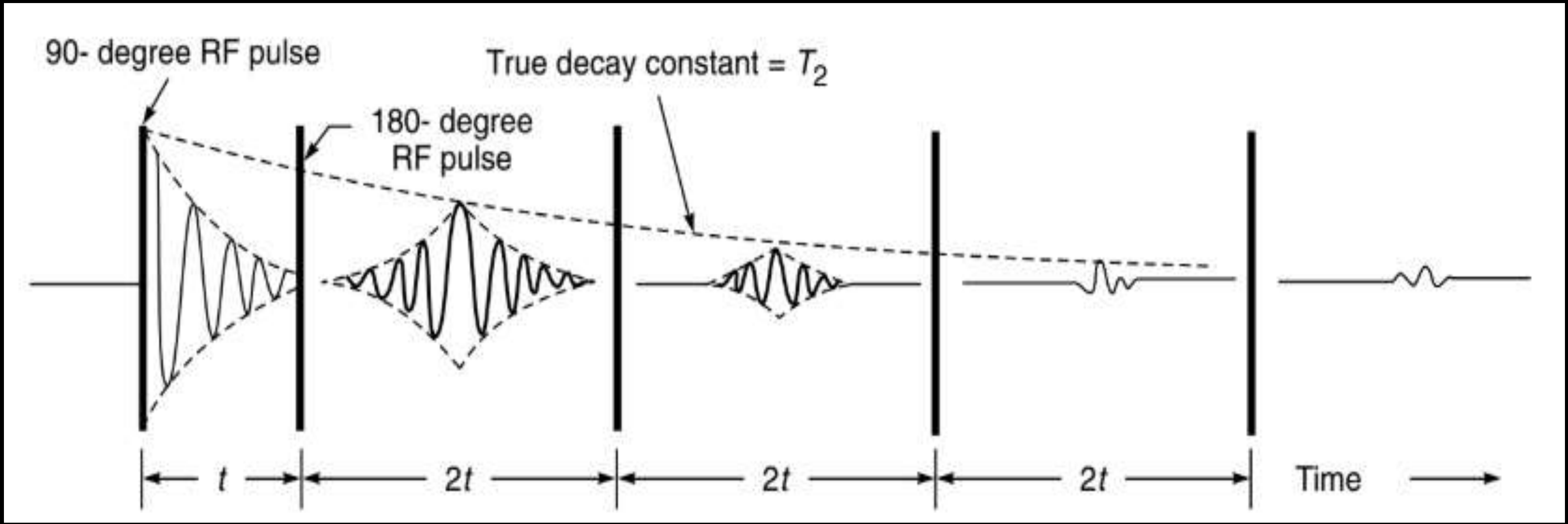


PROTOCOLS IN MR IMAGING

- T_1 weighted imaging
- T_2 weighted imaging
- Proton Density weighted imaging
- Diffusion weighted imaging
- Perfusion imaging

SPIN ECHO IMAGING

- Spin-echo Imaging Technique: The spin-echo imaging technique is useful for creating images that are primarily or solely dependent on T2.
- Figure illustrates the pulse sequence used in this technique. It consists in applying a 90° pulse to rotate the magnetization to the X-Y plane.
- This is followed by periodically flipping X-Y magnetization through 180° by means of a 180° RF pulse.



Parameters

For T1 Weighting

- ✓TR 300 -700ms
- ✓Effective TE minimum
- ✓Turbo factor 2-8 .

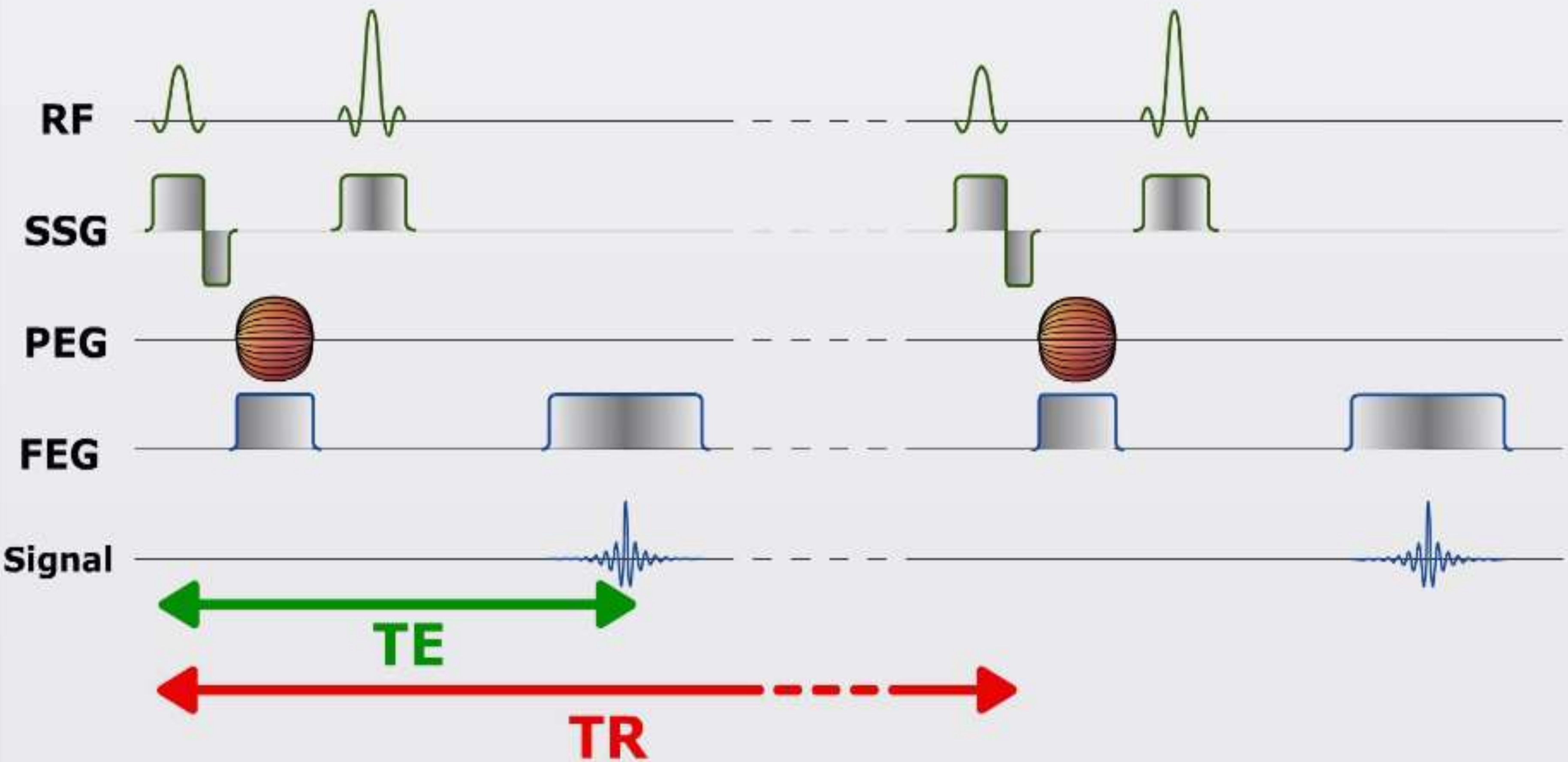
For PD weighting

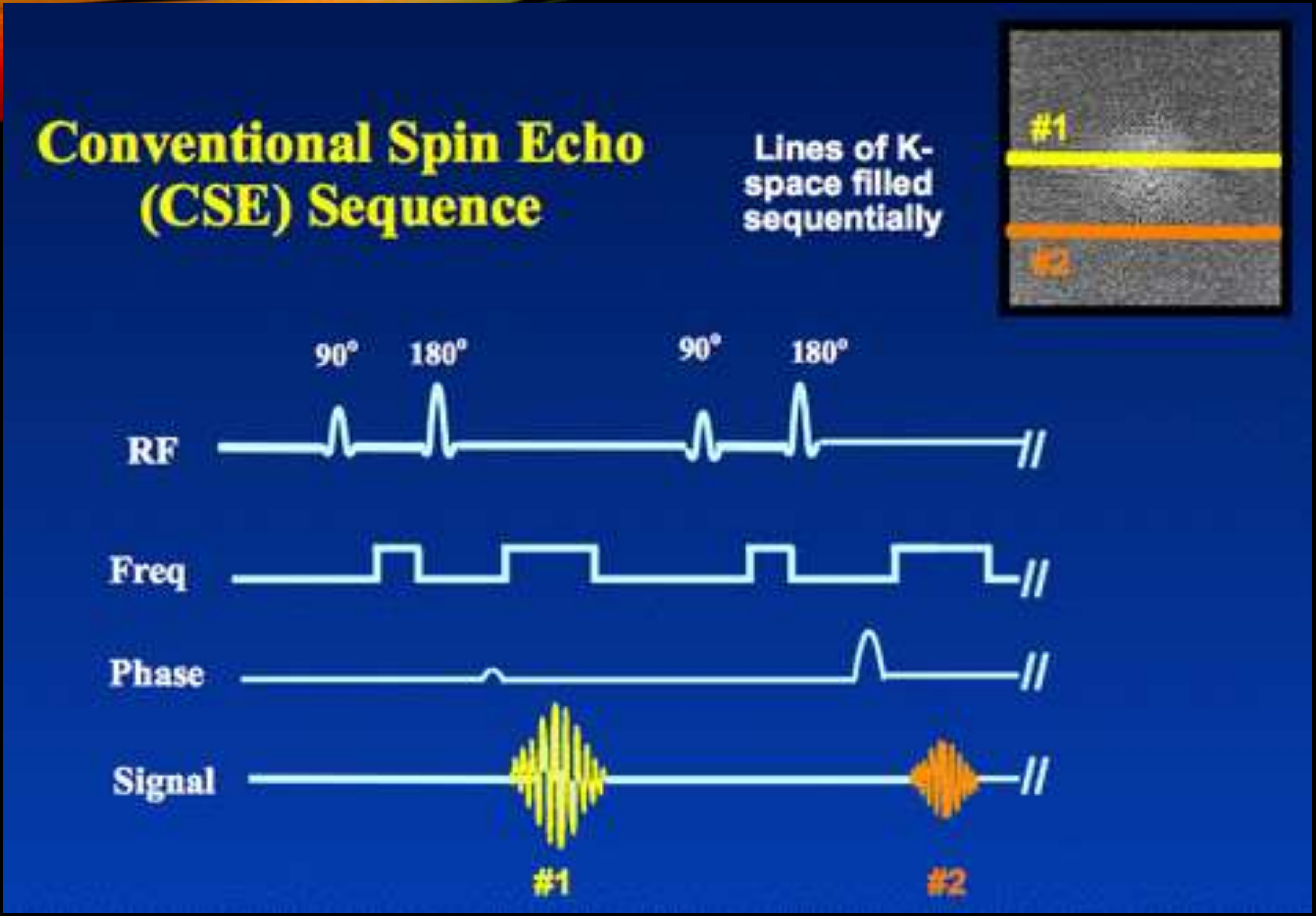
- ✓ TR 3000-10000ms (depending on required slice number)
- ✓ effective TE minimum
- ✓ turbo factor 2-8 .

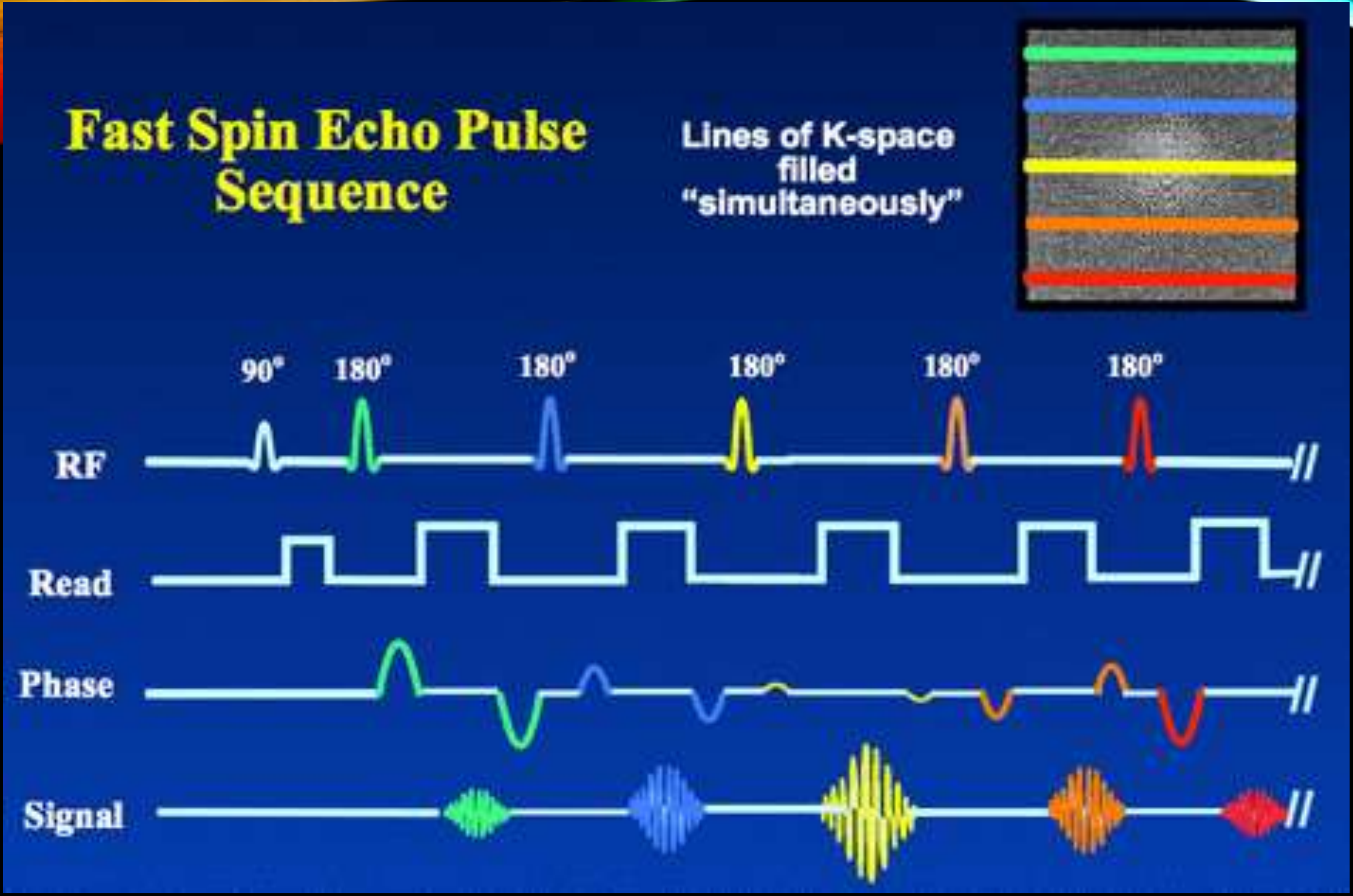
For T2 weighting

- ✓TR 3000-10000ms (depending on required slice number)
- ✓Effective TE 80-140ms
- ✓turbo factor 12-30

Spin Echo



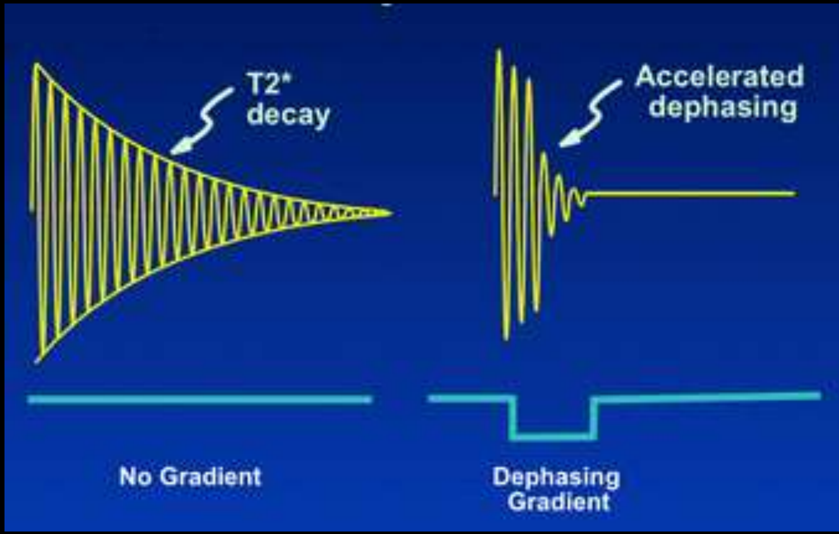

Conventional Spin Echo (CSE) Sequence
Lines of K-space filled sequentially
#1
#2
90°
180°
90°
180°
RF
Freq
Phase
Signal
#1
#2


Fast Spin Echo Pulse Sequence
Lines of K-space filled "simultaneously"
90°
180°
180°
180°
180°
180°
RF
Read
Phase
Signal

GRE SEQUENCE

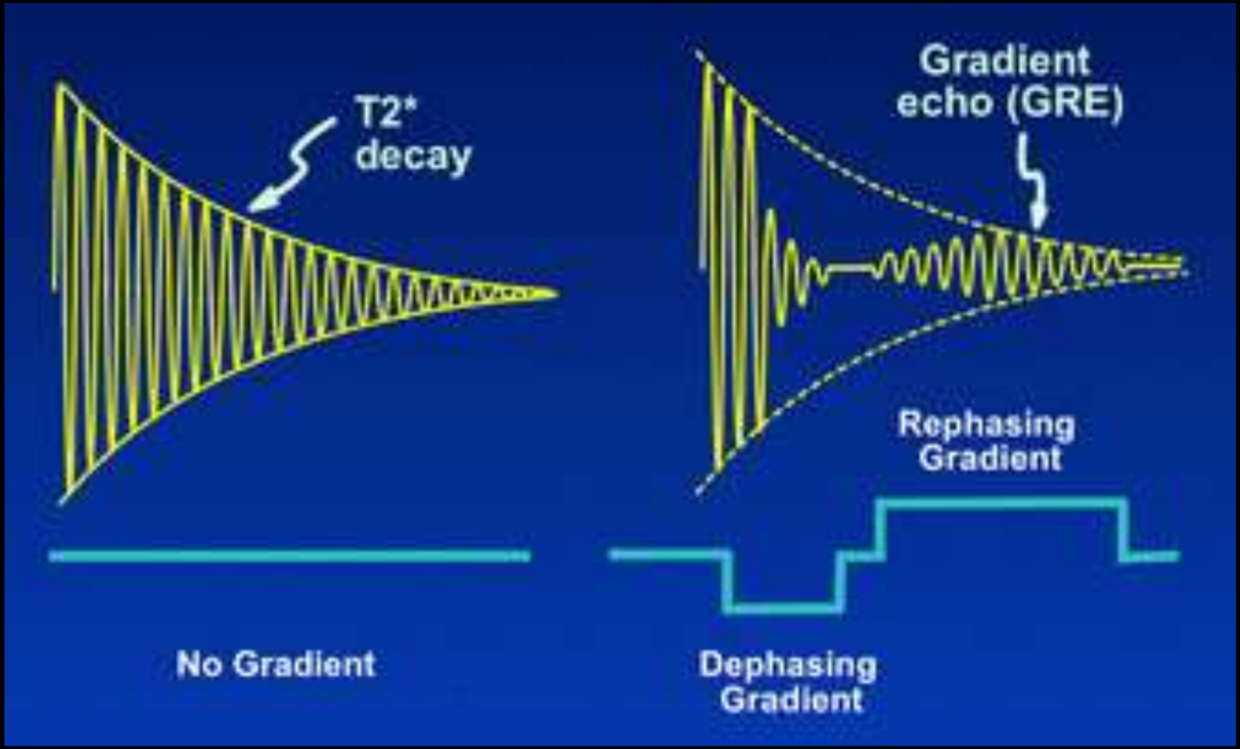
- Gradient echoes are also referred to as ***gradient-recalled echoes*** or ***field echoes***

- A ***gradient echo (GRE)*** is simply a clever manipulation of the FID signal that begins by applying an external ***dephasing gradient*** field across the specimen or tissue



First step in GRE formation –
A gradient is applied across the specimen, resulting in accelerated dephasing and squelching of the FID

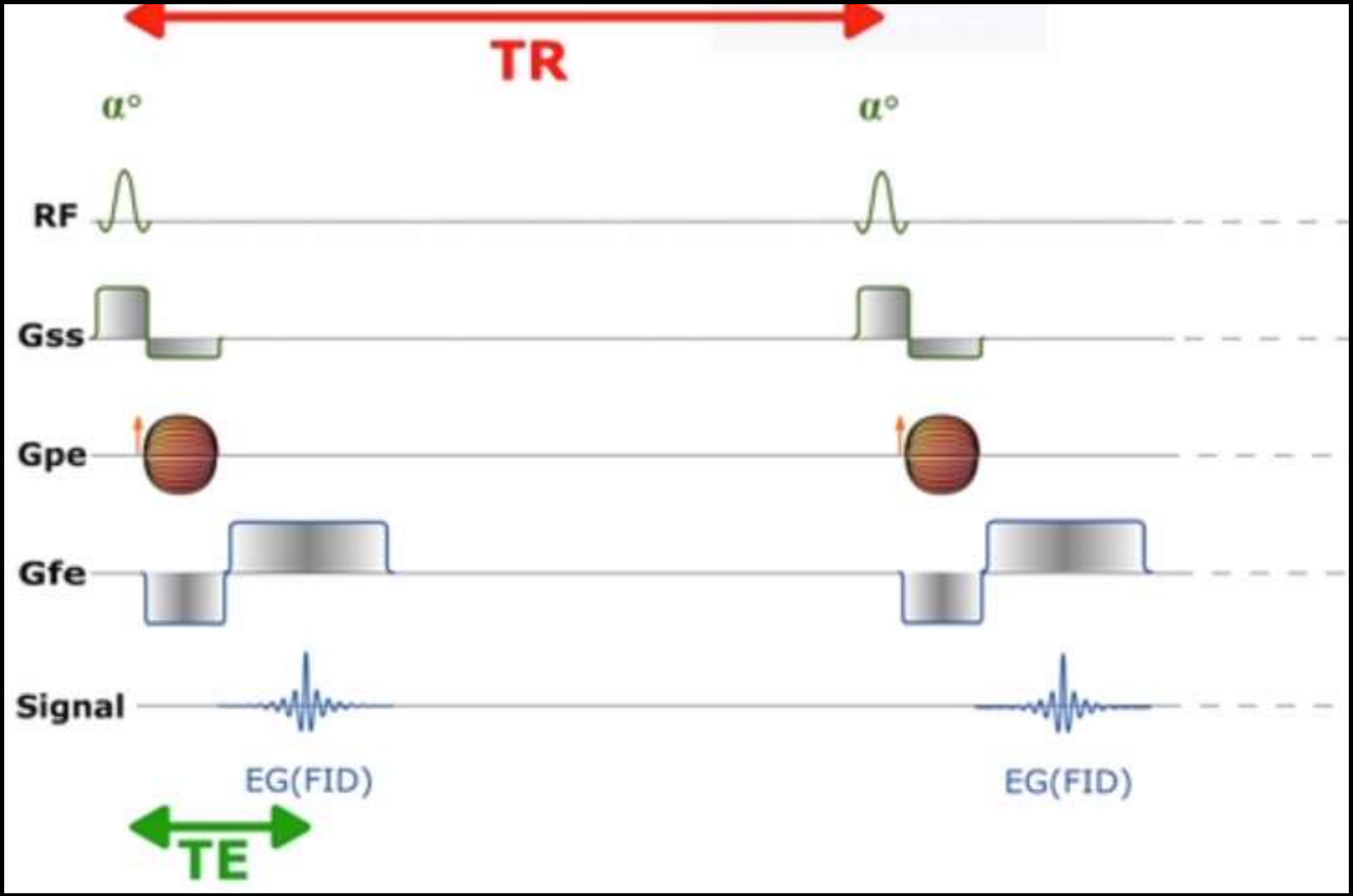
GRE SEQUENCE



Second step in GRE formation - A rephasing gradient is applied (opposite in polarity to the dephasing gradient).

This reverses the phase shifts induced by the dephasing gradient and resurrects the FID as GRE.

GE SEQUENCE



ANALOGY FOR GRE SEQUENCE

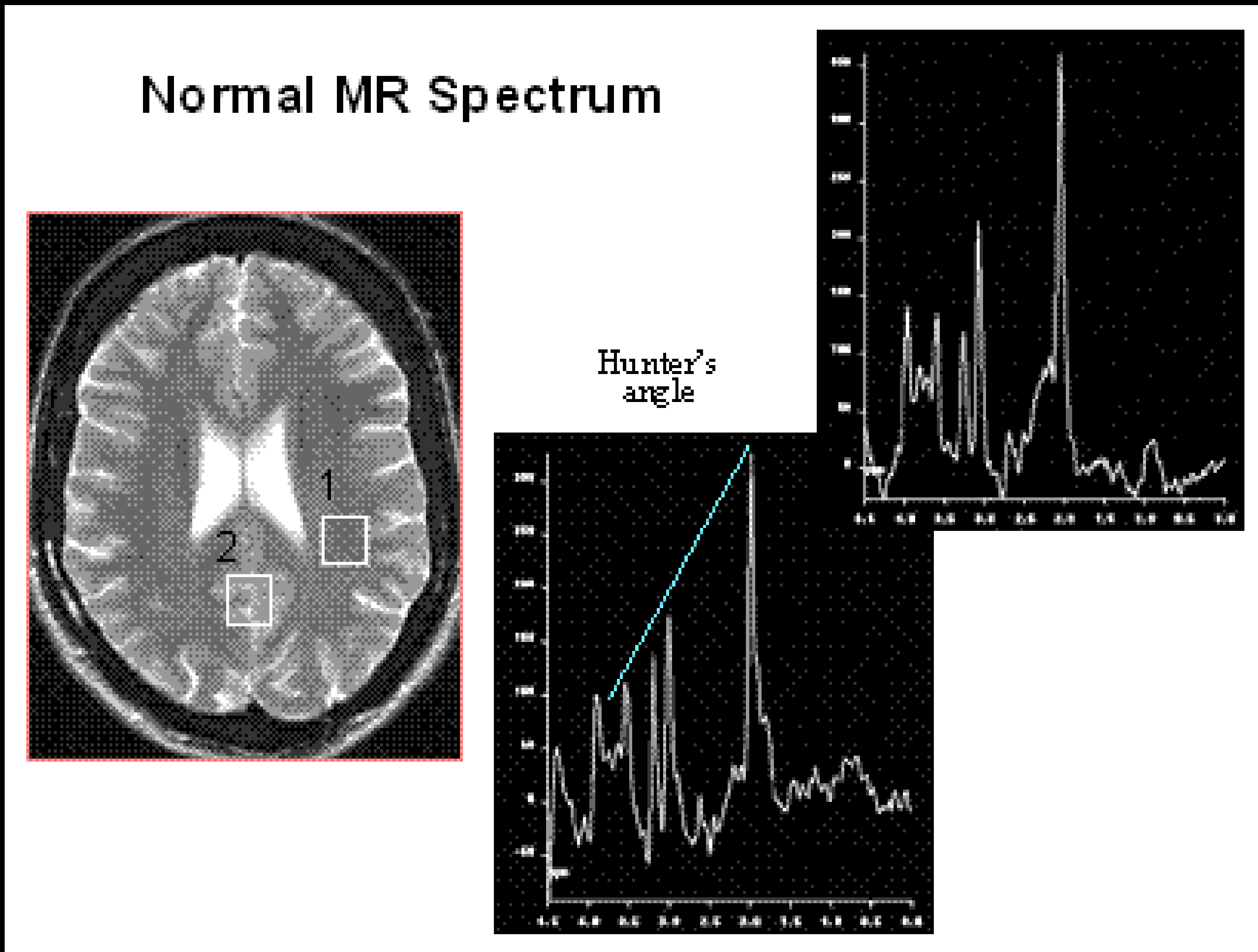
The fast hare represents spins precessing rapidly (and accumulating phase) by virtue of their location in a stronger portion of the gradient;
the tortoise represents more slowly precessing spins in a weaker part of the gradient.

MR SPECTROSCOPY

- Non-invasive physiologic imaging of brain that measures relative levels of various tissue metabolites.

- Used to complement MRI in characterization of various tissues.

MRSI

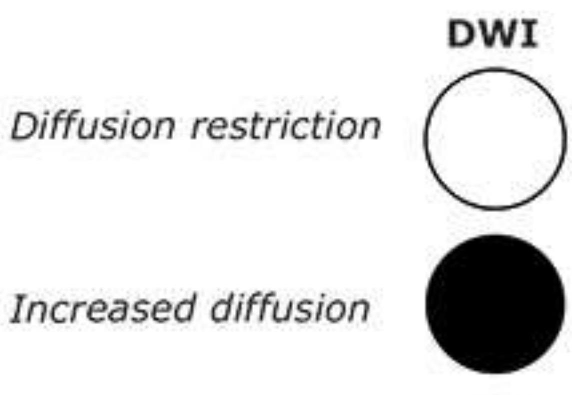


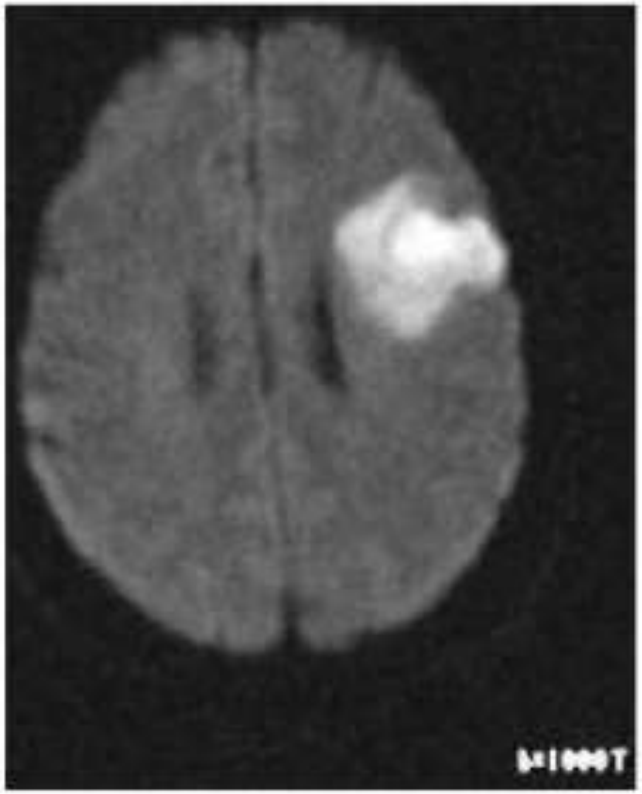
DIFFUSION WEIGHTED IMAGING

- Free water diffusion in the images is Dark (Normal)
- Acute stroke, cytotoxic edema causes decreased rate of water diffusion within the tissue i.e. **Restricted Diffusion** (due to inactivation of Na K Pump)
- Increased intracellular water causes cell swelling

DWI IMAGE

- Areas of restricted diffusion are BRIGHT.
- Restricted diffusion occurs in
  - **Cytotoxic edema**
  - **Ischemia (within minutes)**
  - **Abscess**





PERFUSION MRI

- Promising tool in assessing stroke, brain tumors, and patients with neurodegenerative diseases
- Perfusion MRI is based on the analysis of the contrast enhancement of MRI images after a peripheral injection of a contrast agent
- Three methods are commonly used
    - Dynamic Susceptibility Contrast (DSC)
    - Dynamic Contrast Enhanced (DCE)
    - Arterial Spin Labeling (ASL)
- DSC and DCE require intravenous bolus administration of gadolinium, while ASL is performed without exogenous contrast.

http://mriquestions.com/

PERFUSION MRI

- Dynamic Susceptibility Contrast (DSC): most widely used; rapid T2* imaging after Gd bolus
- Dynamic Contrast Enhanced (DCE): T1-imaging after Gd bolus; measures vascular permeability
- Arterial Spin Labeling (ASL): uses magnetically tagged endogenous water (not Gd) as tracer

https://youtu.be/23h5bNarFvM

BIOLOGICAL EFFECTS OF NMR IMAGING

The three aspects of NMR imaging which could cause potential health hazard are:

1. Heating due to the rf power: Katinis (1982) reports that a temperature increase produced in the head of NMR imaging would be about 0.3°C. This does not seem likely to pose a problem.
2. Static magnetic field: Although no significant effects of the static field with the level used in NMR are known, Pastakia (1978) mentions about the possible side effects of electromagnetic fields. There could be a slight decrease in cognitive skills, mitotic delay in slime moulds, delayed wound healing and elevated serum triglycerides.
3. Electric current induction due to rapid change in magnetic field: It is believed that oscillating magnetic field gradients may induce electric currents strong enough to cause ventricular fibrillation. However, no damage due to NMR from exposures has been reported (Marx 1980). It is suggested that fields should not vary at a rate faster than 3 tesla/s

Advantages of MRI

The NMR image provides substantial contrast between soft tissues that are nearly identical in existing techniques.

- NMR images that display T1 andT2 properties of tissue provide tremendous contrasts between various soft tissues, contrasts approaching 150% are possible in T1 and T2 images, while contrasts of only a few percent are possible between soft tissues with X-rays.

- Cross-sectional images with any orientation are possible in NMR imaging systems.

- The alternative contrast mechanisms of NMR provide promising possibilities of new diagnostics for pathologies that are difficult or impossible with present techniques.

Advantages of MRI

- NMR imaging parameters are affected by chemical bonding and, therefore, offer potential for physiological imaging
- NMR uses no ionizing radiation and has minimal, if any, hazards for operators of the machines and for patients
- Unlike CT, NMR imaging requires no moving parts, gantries or sophisticated crystal detectors. The system scans by superimposing electrically controlled magnetic fields
- Consequently, scans in any pre-determined orientation are possible.
- With the new techniques being developed, NMR permits imaging of entire three-dimensional volumes simultaneously instead of slice by slice, employed in other imaging systems.

ADVANTAGES OF MRI

1. No ionizing radiation & no short/long-term effects demonstrated
2. Variable thickness, any plane
3. Better contrast resolution & tissue discrimination
4. Various sequences to play with to characterise the abnormal tissue
5. Many details without I.V contrast

DISADVANTAGES

The three aspects of NMR imaging which could cause potential health hazard are:

(i) Heating due to the rf power: Katinis (1982) reports that a temperature increase produced in the head of NMR imaging would be about 0.3°C. This does not seem likely to pose a problem

(ii) Static magnetic field: Although no significant effects of the static field with the level used Pastakia (1978) mentions about the possible side effects of electro- magnetic fields. There could be a slight decrease in cognitive skills, mitotic delay in slime moulds, delayed wound healing and elevated serum triglycerides

(iii) Electric current induction due to rapid change in magnetic field: It is believed that oscillating magnetic field gradients may induce electric currents strong enough to cause ventricular fibrillation. However, no damage due to NMR from exposures has been reported (Marx 1980). It is suggested that fields should not vary at a rate faster than 3 tesla/s.

DISADVANTAGES OF MRI

- Time consuming
- Not easily available (long waiting list)
- No on-call service
- Need to tweak sequences as per the clinical questions; hence cannot be generalised

LINKS FOR CONCEPTS

- https://www.youtube.com/watch?v=5sphPrRDdek
- Animation on kspace

- https://www.youtube.com/watch?v=83cgR7m8lT8
- Gradient field and coils

- https://www.youtube.com/watch?v=mOt2FeGHjaY
- T1 and T2 image

- https://www.youtube.com/watch?v=DYXEGY-X1n8&pbjreload=10
- T1 AND T2 IMAGES DETAILS

- https://www.imaios.com/en/e-Courses/e-MRI/MRI-Sequences/gradient-echo
- Flip angle

- https://www.imaios.com/en/e-Courses/e-MRI/The-Physics-behind-it-all/K-space
- MR image formation