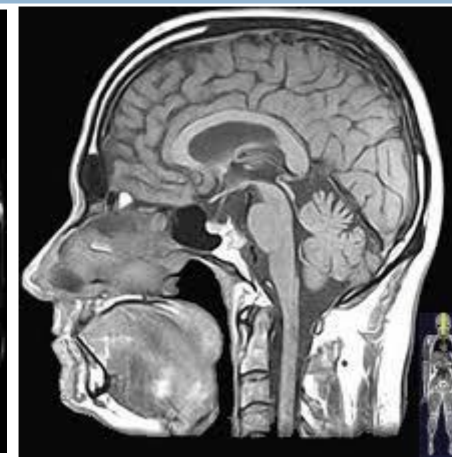
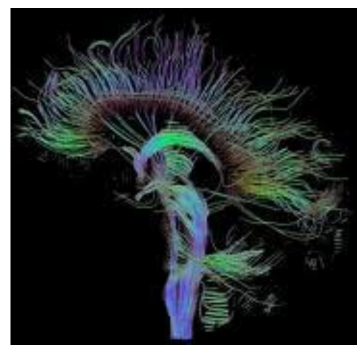
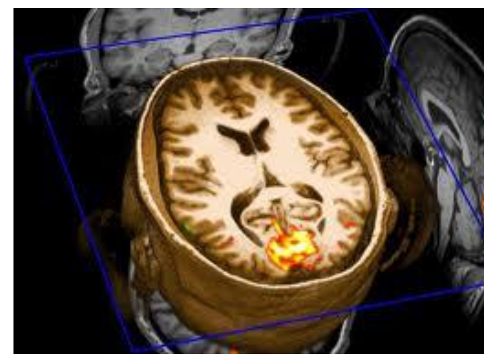
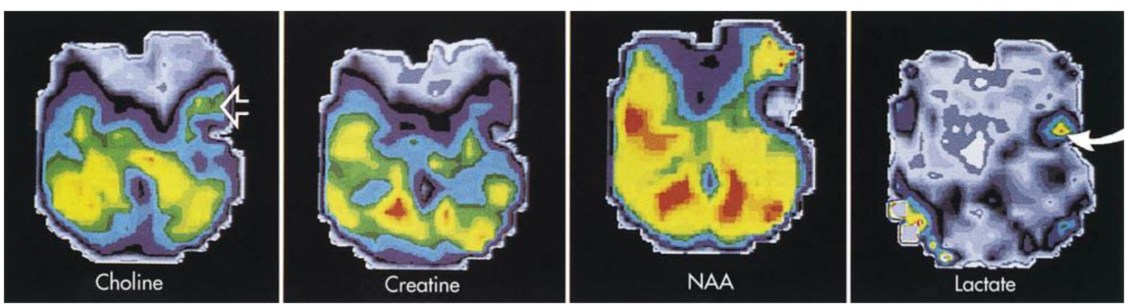
CAIRO UNIVERSITY

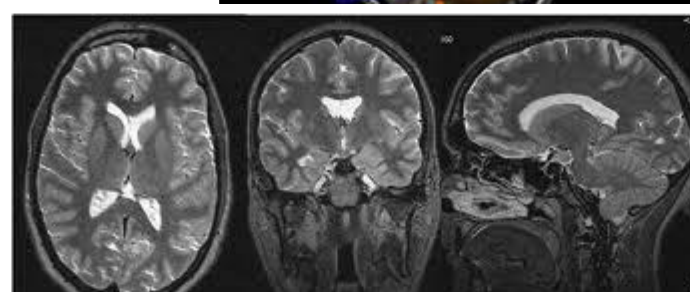
MAGNETIC RESONANCE IMAGING

Prof. Yasser Mostafa Kadah

Magnetic Resonance Imaging

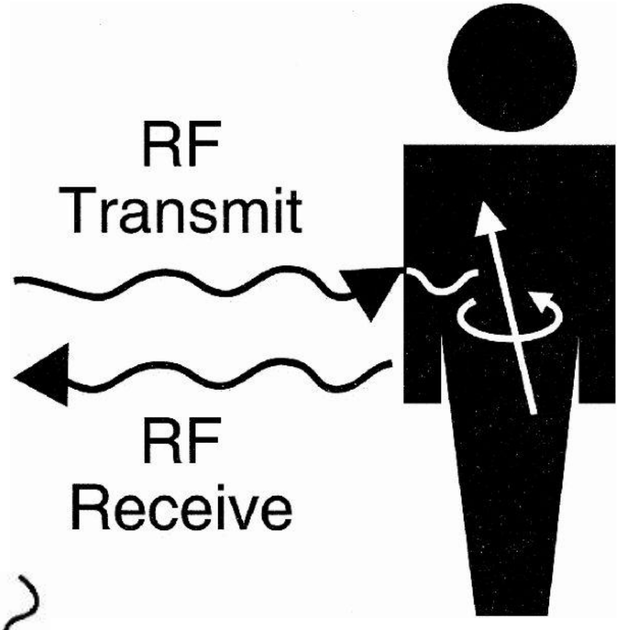
- Anatomy
- Physiology (function)
- Angiography
- Diffusion
- Perfusion
- Spectroscopy

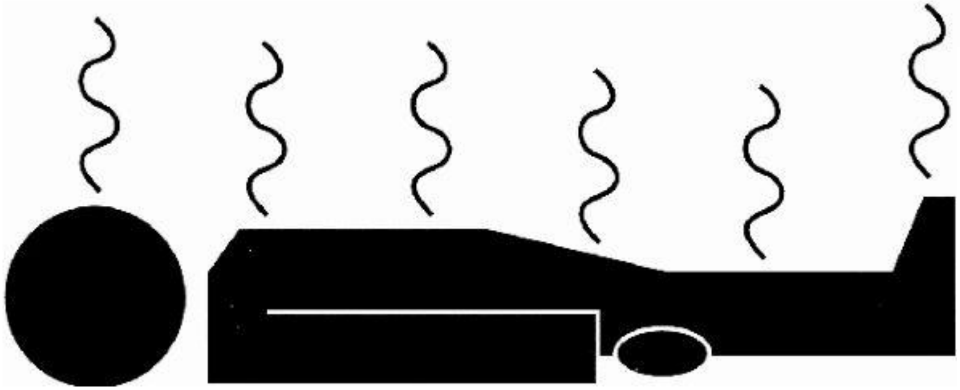








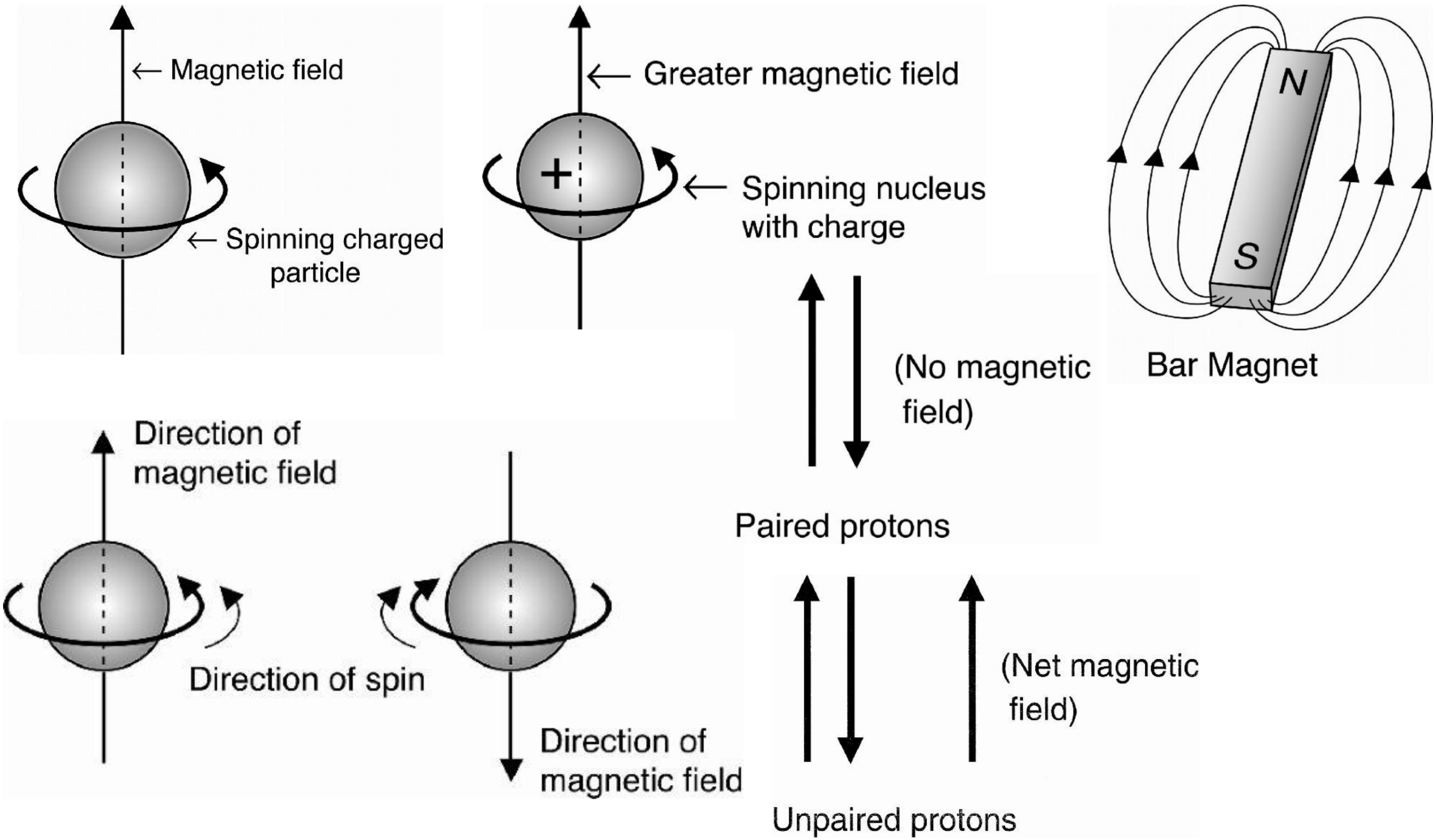
Steps to Perform MR Imaging

- **M**: Magnetic Field
  - Patient is placed inside magnet
- **R**: Radio-Frequency (RF) Pulse
  - RF pulse is applied
- **R**: Relaxation
  - After RF application, signal is collected from relaxation



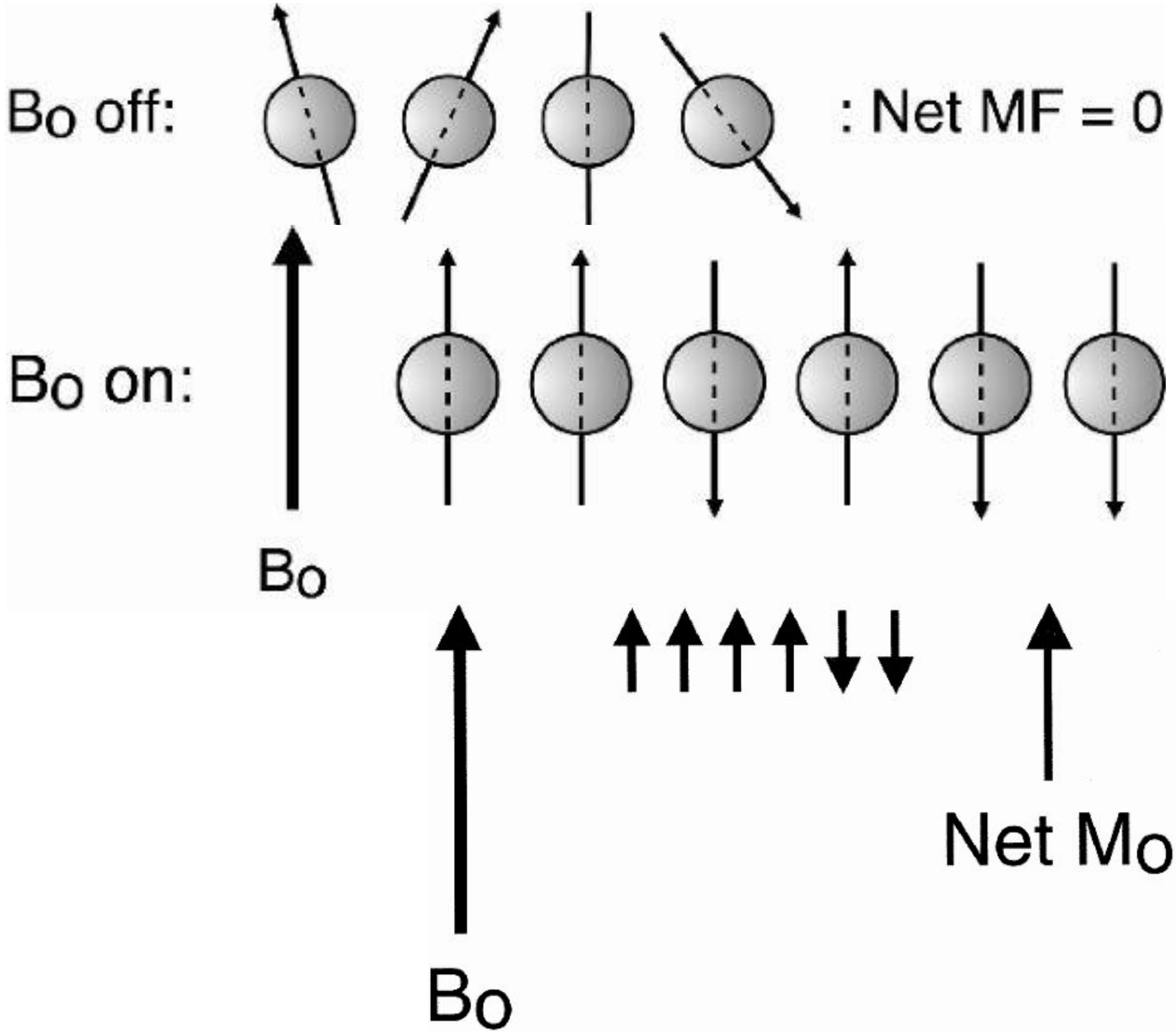


Basic Physics



B0 Field

- The external magnetic field is denoted B0 (read as "B-zero")
- In MRI, B0 is on the order of 1 Tesla (1T)
    - One Tesla is equal to 10,000 Gauss.



Net Magnetization Magnitude

- Follow Boltzmann distribution $\boxed{e^{-(U/k_BT)} = e^{\gamma m\hbar B/k_BT}}$

$$\boxed{\langle \mu_z \rangle = \frac{\gamma\hbar \sum\limits_{m=-I}^{I} m\, e^{\gamma m\hbar B/k_BT}}{\sum\limits_{m=-I}^{I} e^{\gamma m\hbar B/k_BT}}.}$$

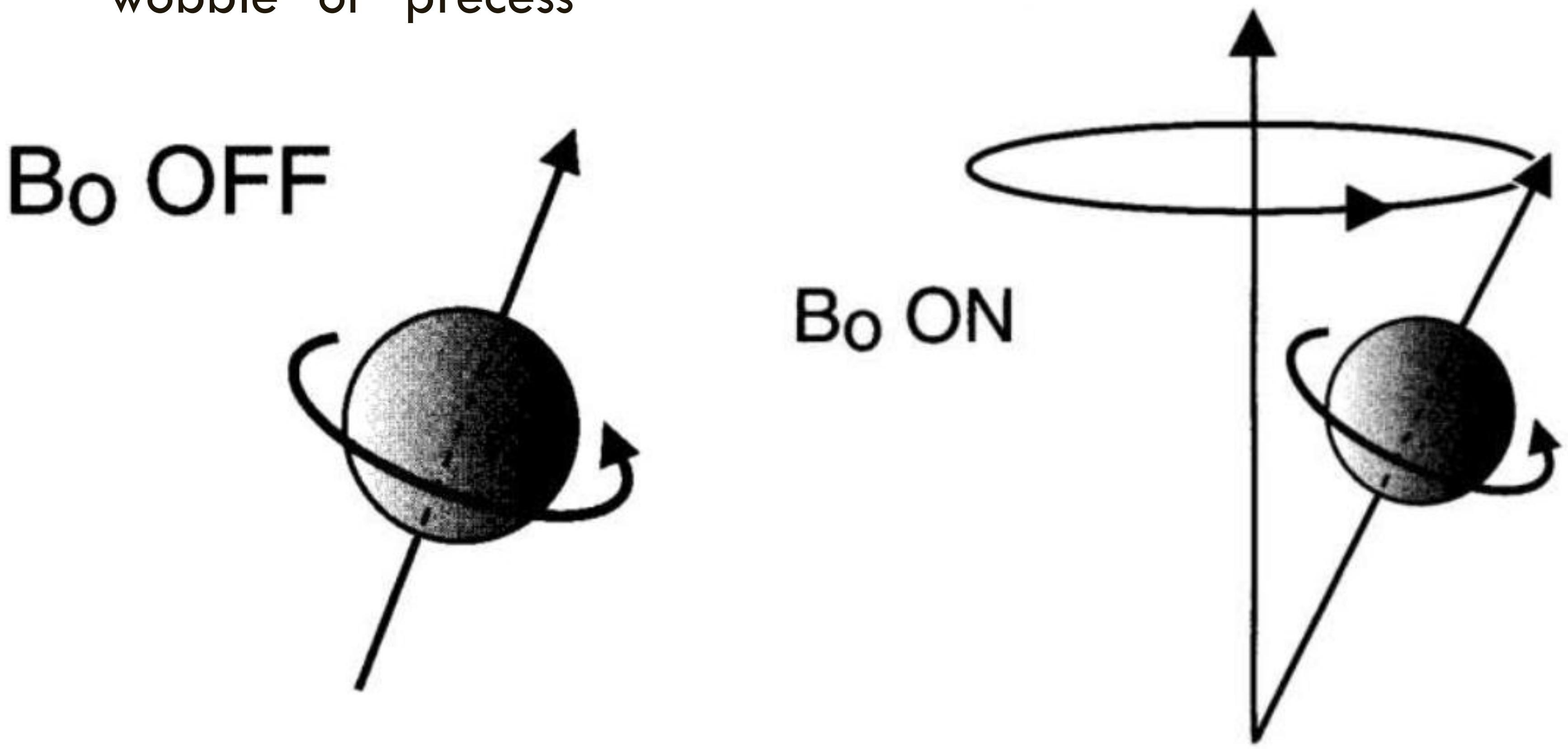
- At room temperature, $\gamma I\hbar B/k_BT \ll 1$

$$\boxed{M_z = N\langle \mu_z \rangle = \frac{N\gamma^2\hbar^2 I(I+1)}{3k_BT}B.}$$

⇨ M_z is proportional to the applied field B0

Precession

- When a proton is placed in a large magnetic field, it begins to "wobble" or "precess"



Larmor Equation

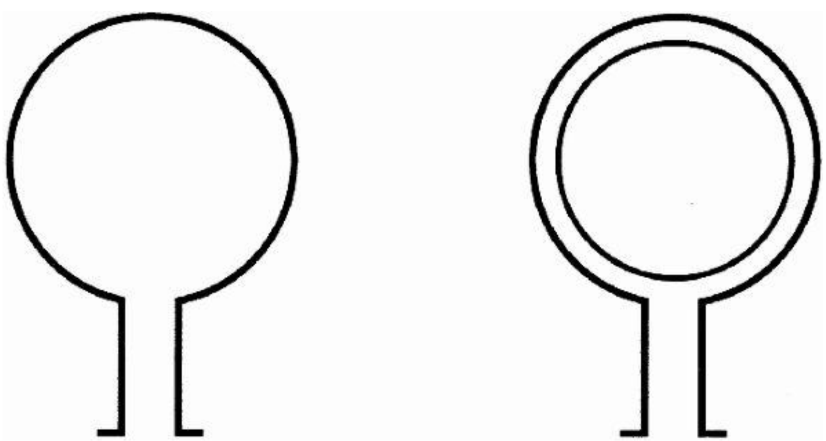
- The rate at which proton precesses around external magnetic field is given by:

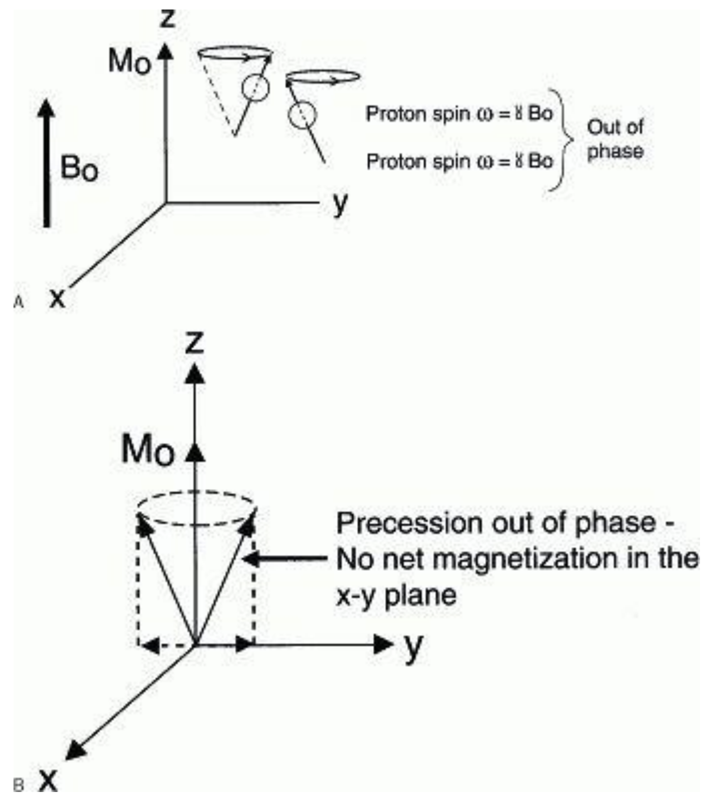
$$\boxed{\omega = \gamma B_0}$$

Nucleus	Spin Quantum Number (S)	Gyromagnetic Ratio* (MHz/T)
^{1}H	1/2	42.6
^{19}F	1/2	40.0
^{23}Na	3/2	11.3
^{13}C	1/2	10.7
^{17}O	5/2	5.8

Problem in MRI Signal Acquisition

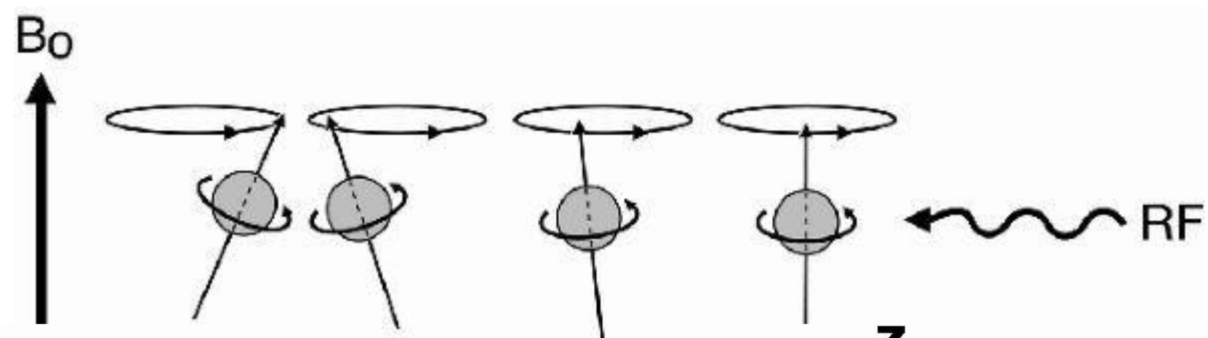
- B0 field is much larger than tissue net magnetization
  - Impossible to measure net magnetization in the z-direction
  - Need to look at component on x-y plane
  - Problem: x-y components cancel out
- Measured using pick-up coils

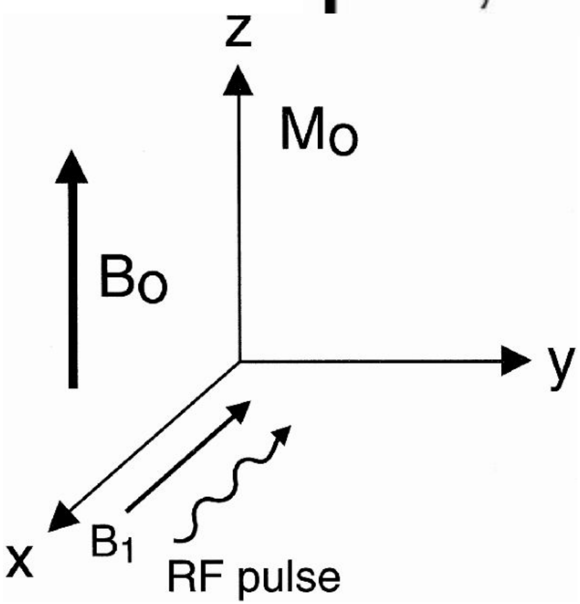


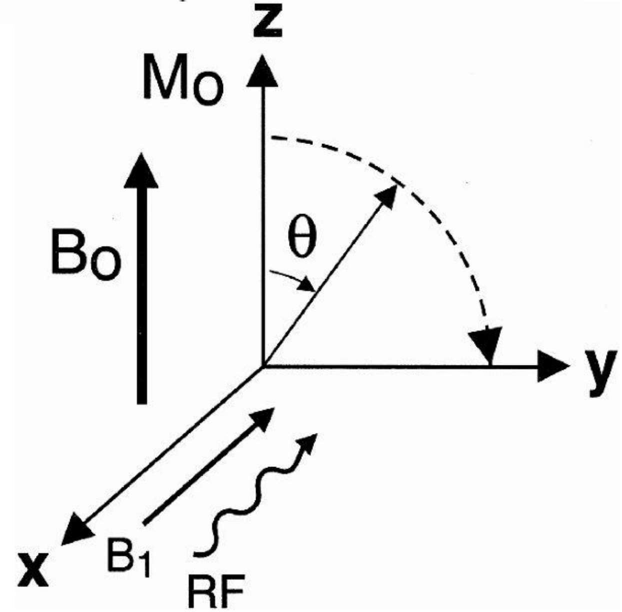


RF Pulse

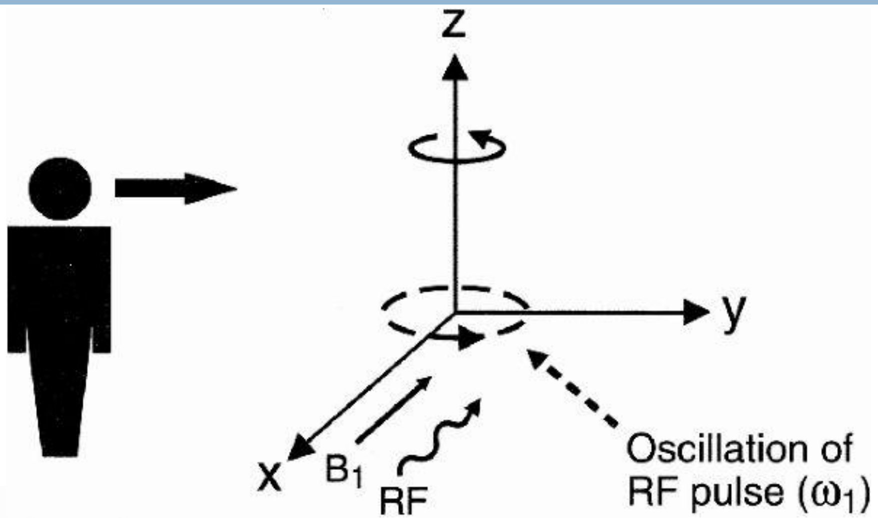
- Idea: Sending RF radiation at Larmor frequency to flip net magnetization to x-y plane

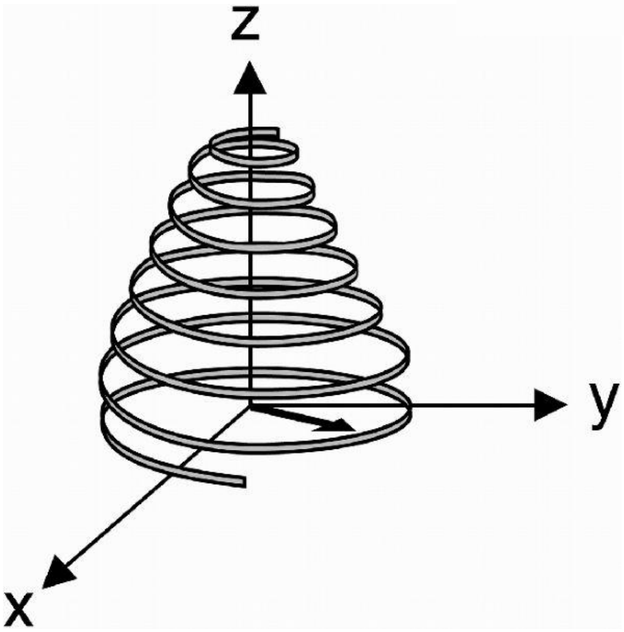


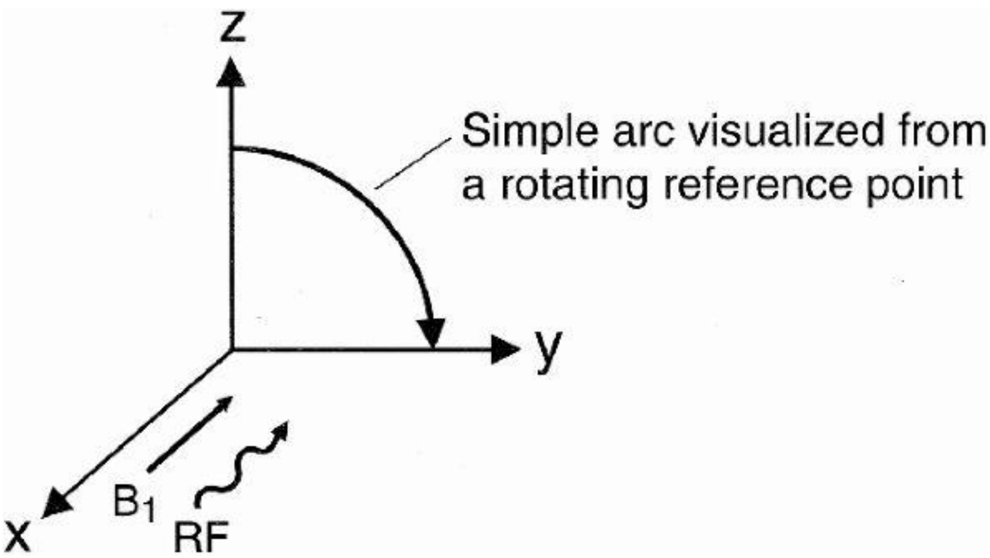




Rotating Frame of Reference

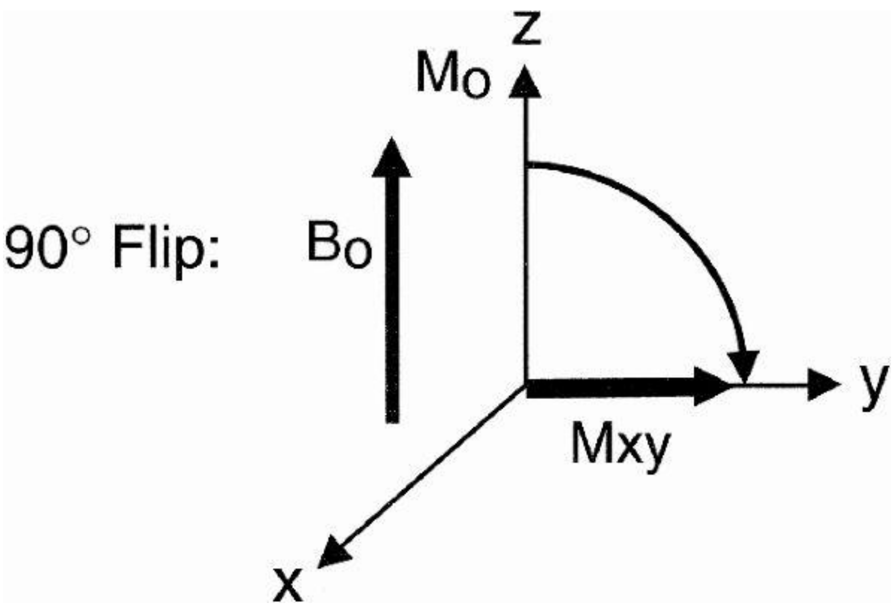


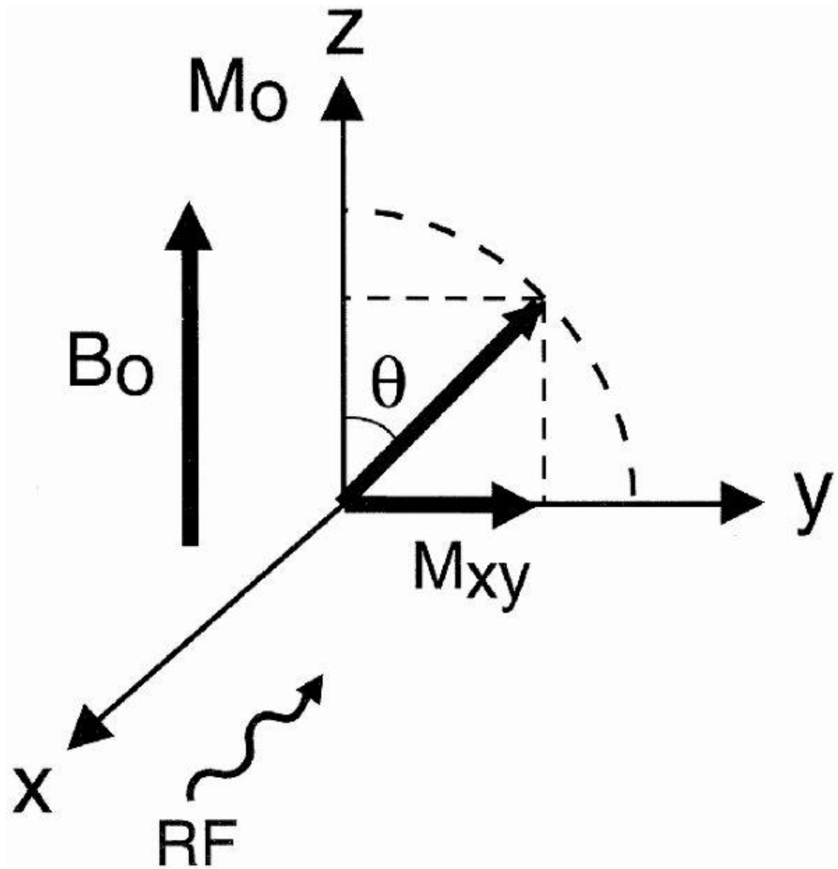




Selection of RF Pulse Flip Angle

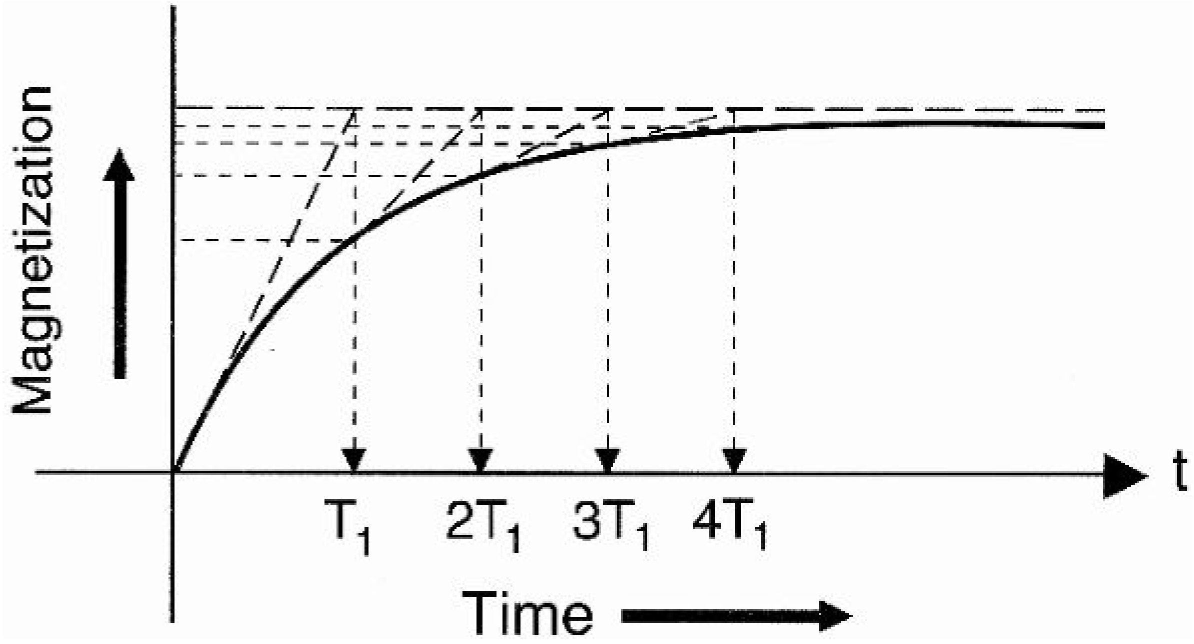
- 90° or 180° or partial flip





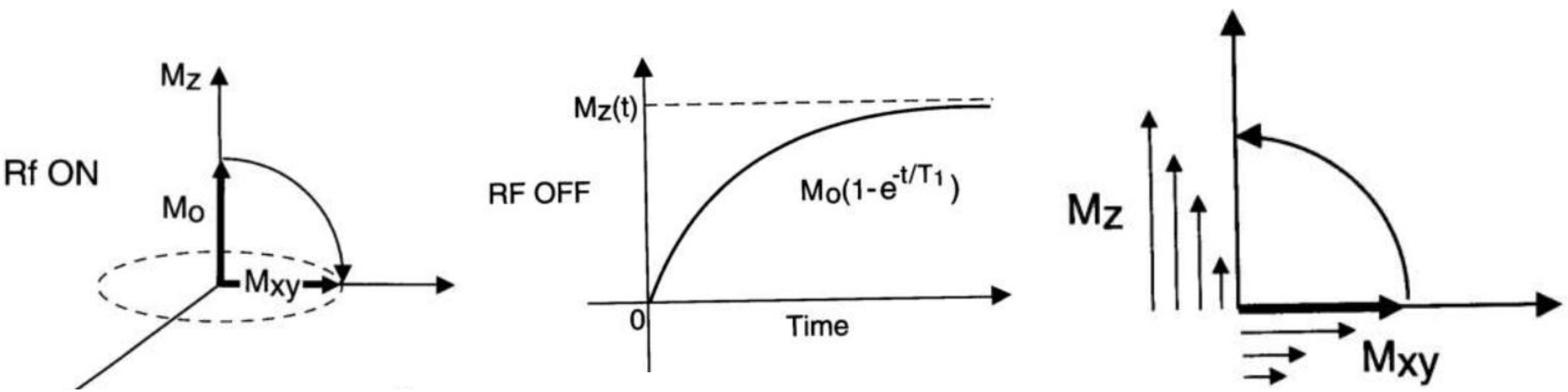
Relaxation

- Relaxation means that the spins are relaxing back into their lowest energy state or back to the equilibrium state
  - Equilibrium by definition is the lowest energy state possible
  - Once the RF pulse is turned off, the protons will have to realign with the axis of the B0 magnetic field and give up all their excess energy



T1 Relaxation

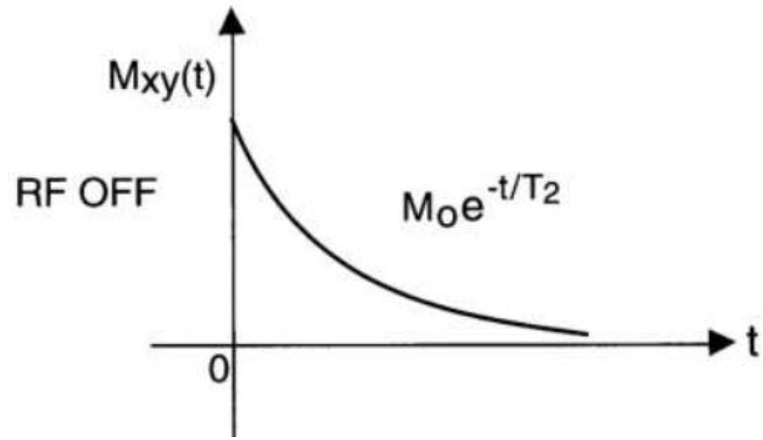
- T1 is called the longitudinal relaxation time because it refers to the time it takes for the spins to realign along the longitudinal (z) axis
- T1 is also called the spin-lattice relaxation time because it refers to the time it takes for the spins to give the energy they obtained from the RF pulse back to the surrounding lattice in order to go back to their equilibrium state.

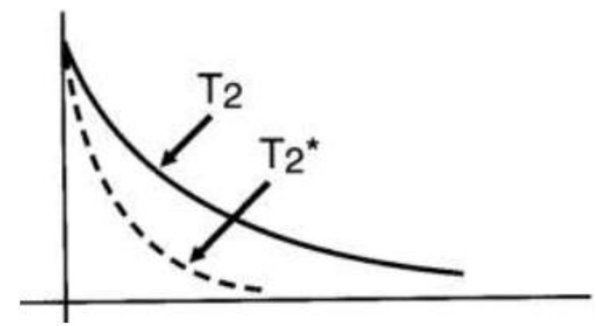


T2 Relaxation

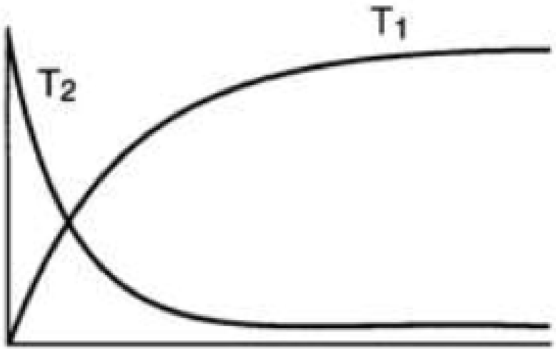
- Dephasing: after the 90° RF pulse is turned off, all spins are in phase; they are all lined up in the same direction and spinning at the same frequency ω_0. There are two phenomena that will make the spins get out of phase: interactions between spins and external field inhomogeneities
- T2 Relaxation
  - Only spin-spin interactions
- T2* Relaxation
  - Both effects

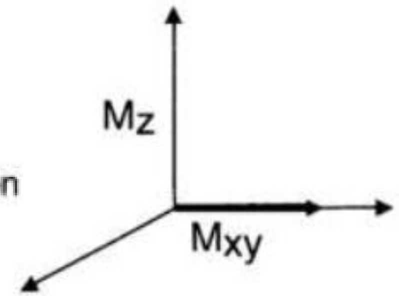
$$\boxed{1/T2^* = 1/T2 + \gamma\Delta B}$$



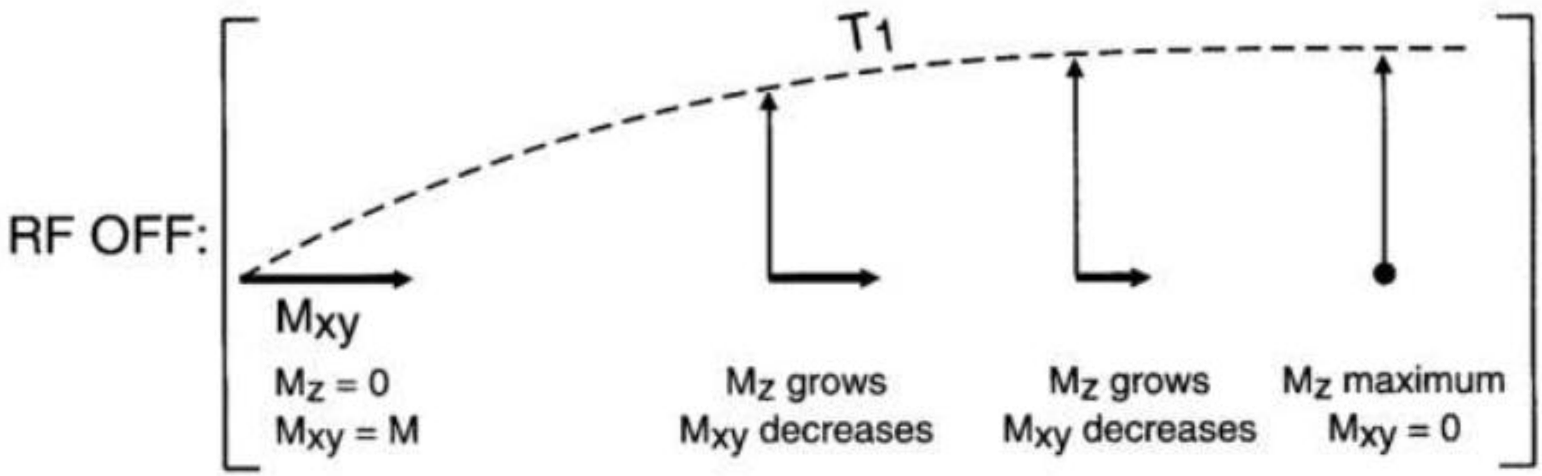


Effect of Both T1 and T2 Relaxations





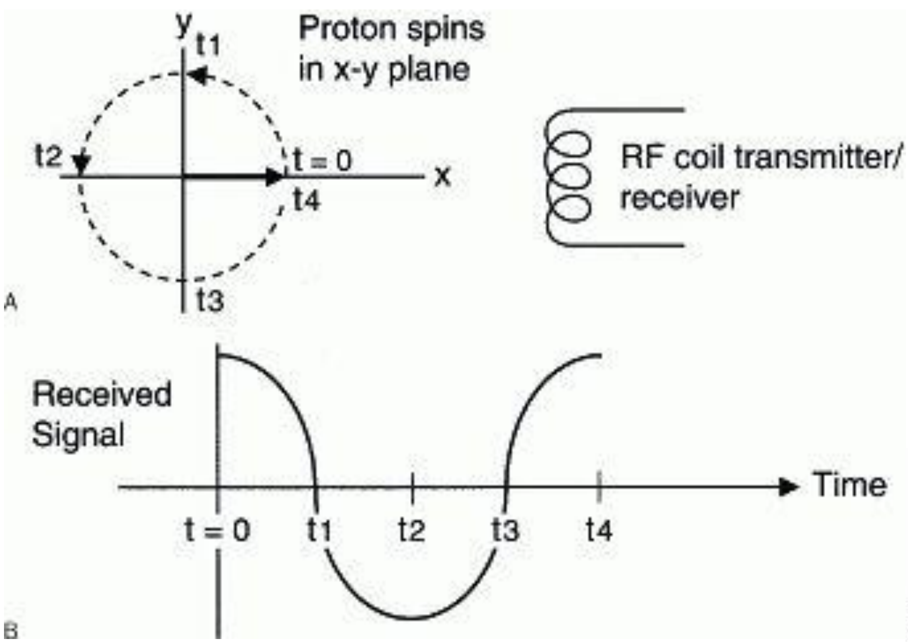
Loss of Mxy
Gain of Mz
Independent processes

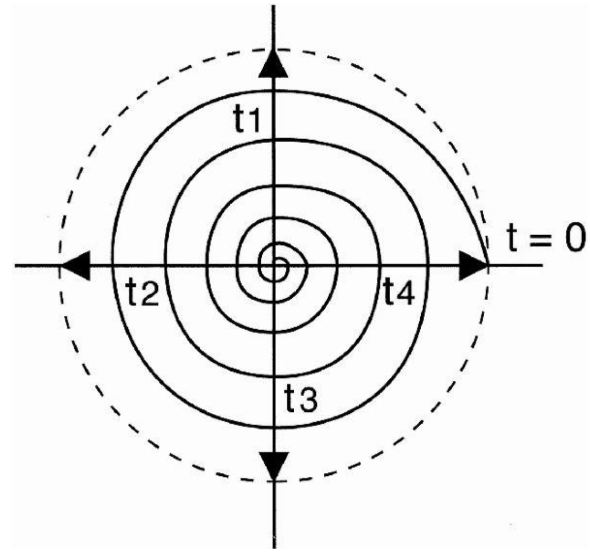


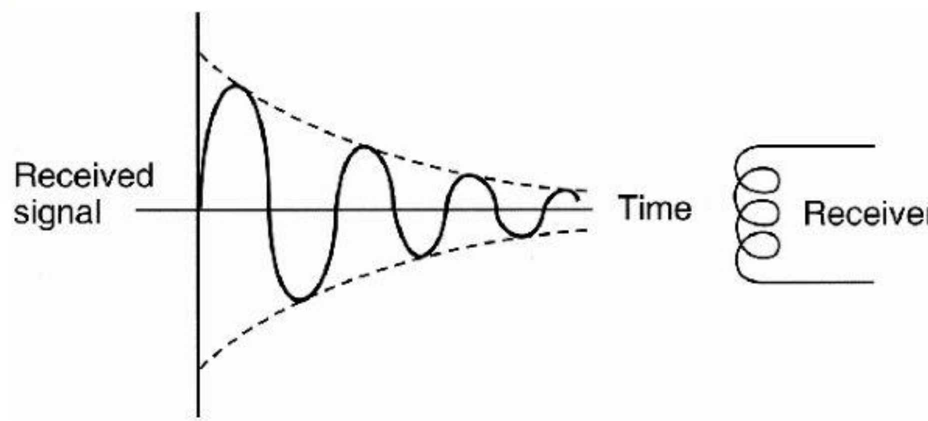
Example Tissue Relaxation Times

Tissue	T_1 (ms)	T_2 (ms)
H_2O	2500	2500
fat	200	100
CSF	2000	300
gray matter	500	100

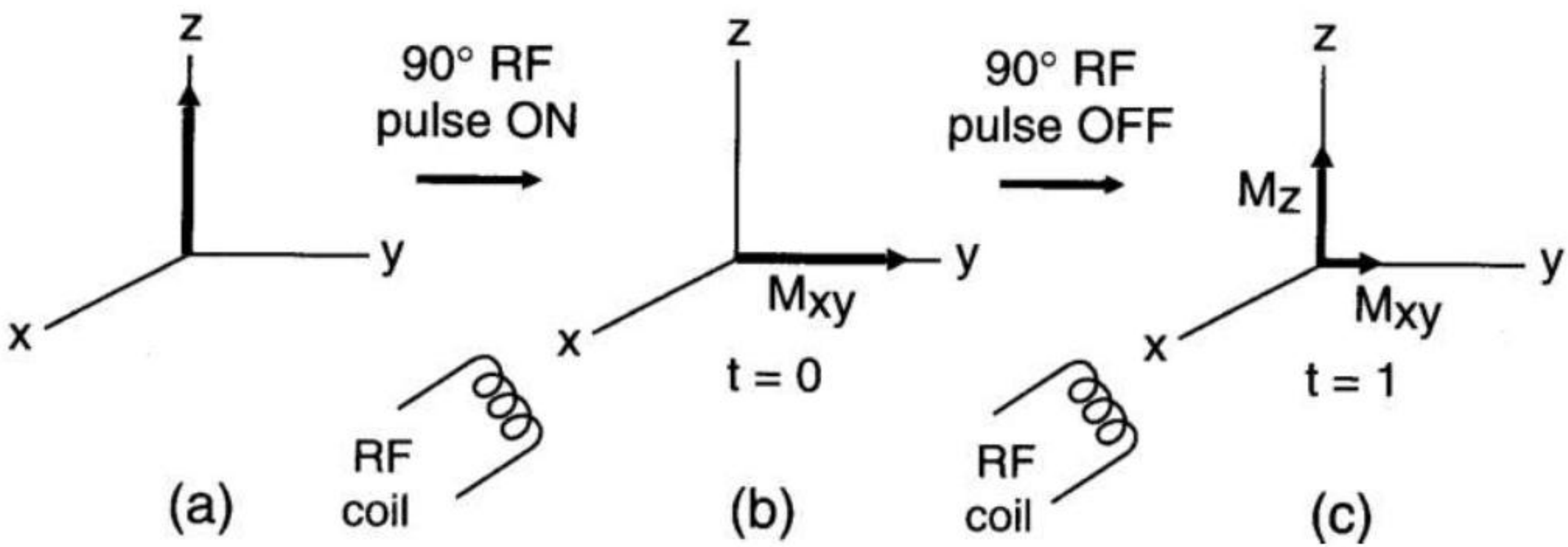
Received Signal: Free Induction Decay (FID)

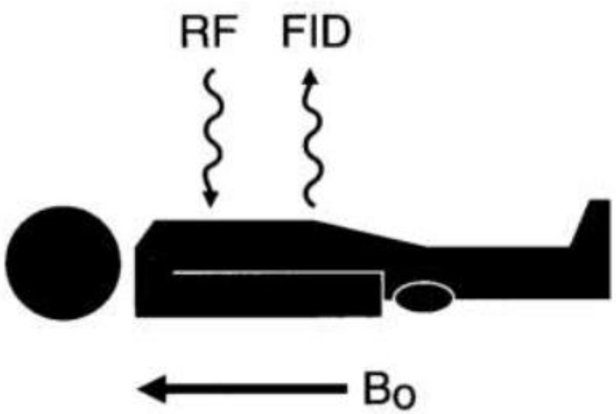


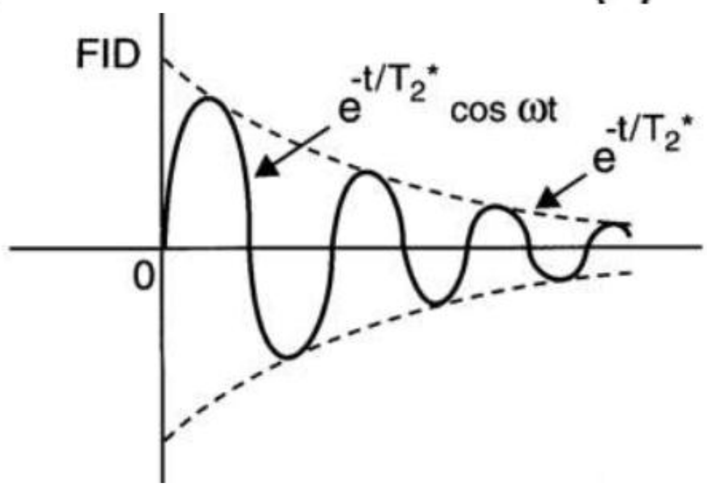




Sequence of Events in MRI

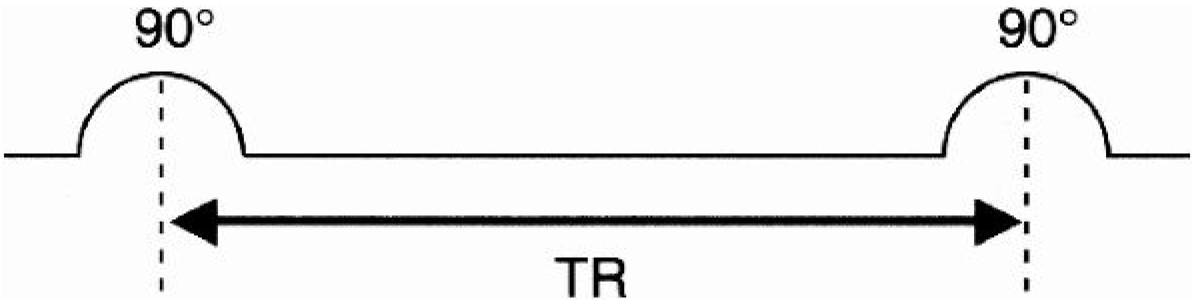






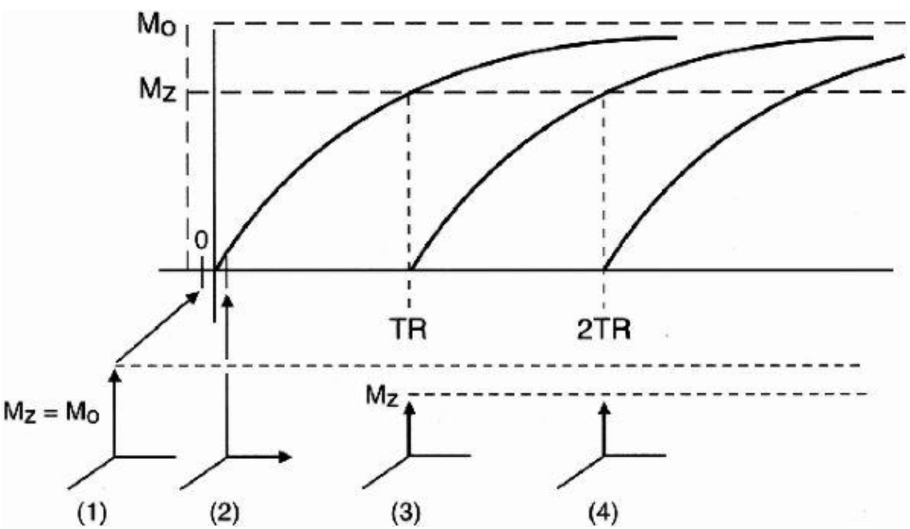
Pulse Repetition Time (TR)

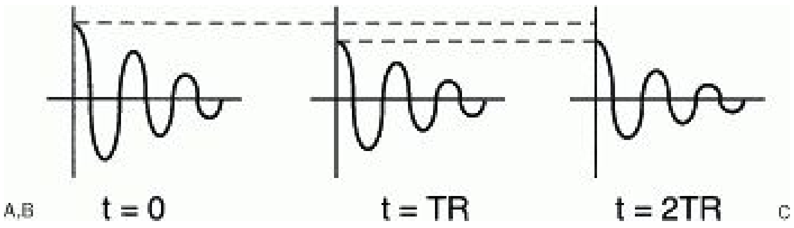
- Distance between successive RF pulses



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

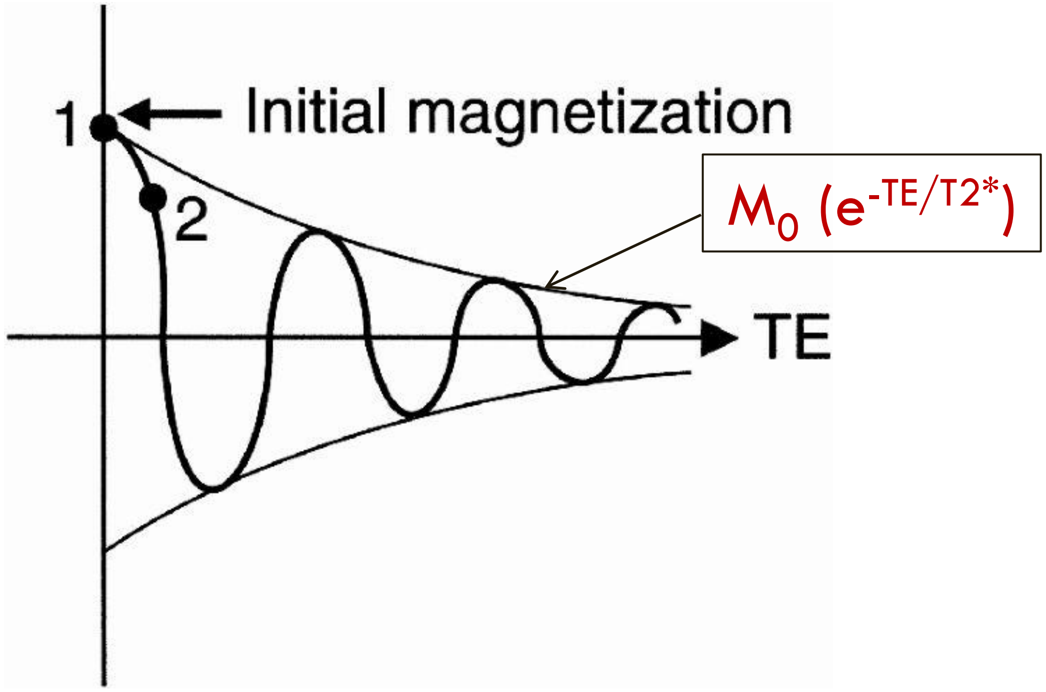
$S \propto N(H)\ (1 - e^{-TR/T1})$





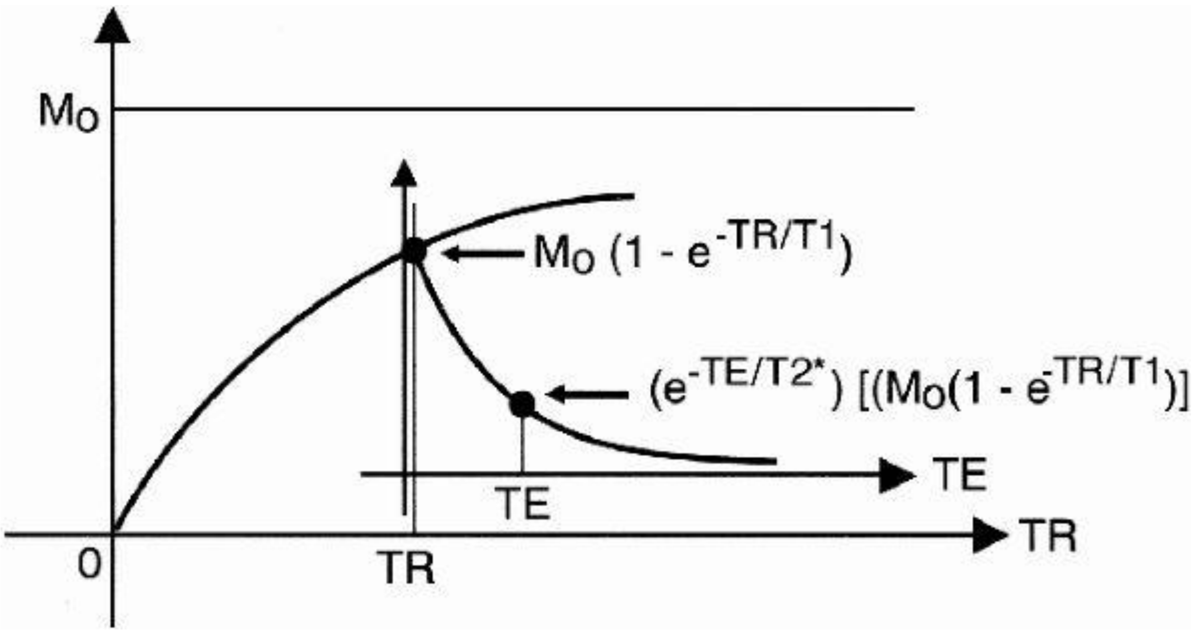
Echo Time or Time to Echo (TE)

- Instead of making the measurement immediately after the RF pulse, we wait a short period of time TE and then make the measurement
  - Time sampling of FID starts



Tissue Contrast

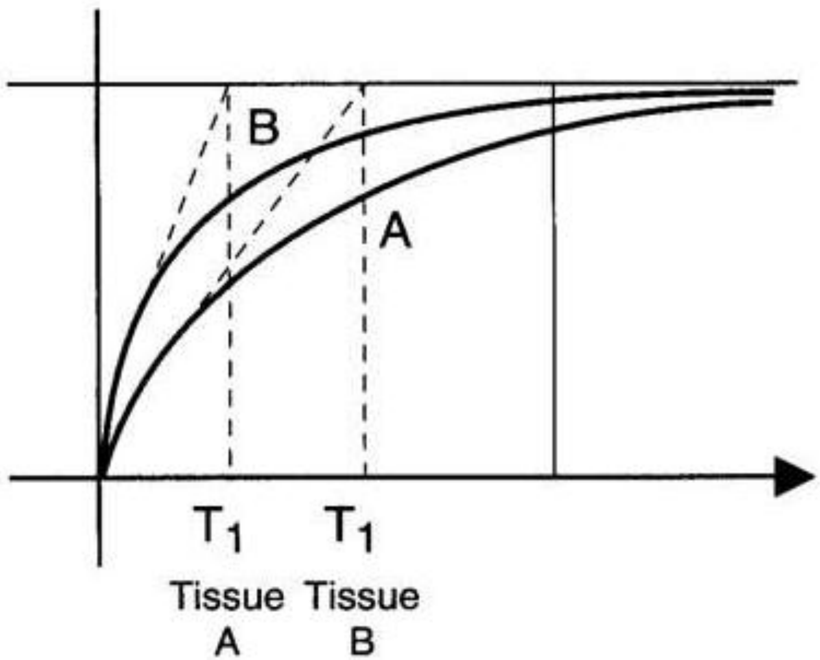
- Now we have to put the two curves together because both T1 recovery and T2 decay processes are occurring simultaneously

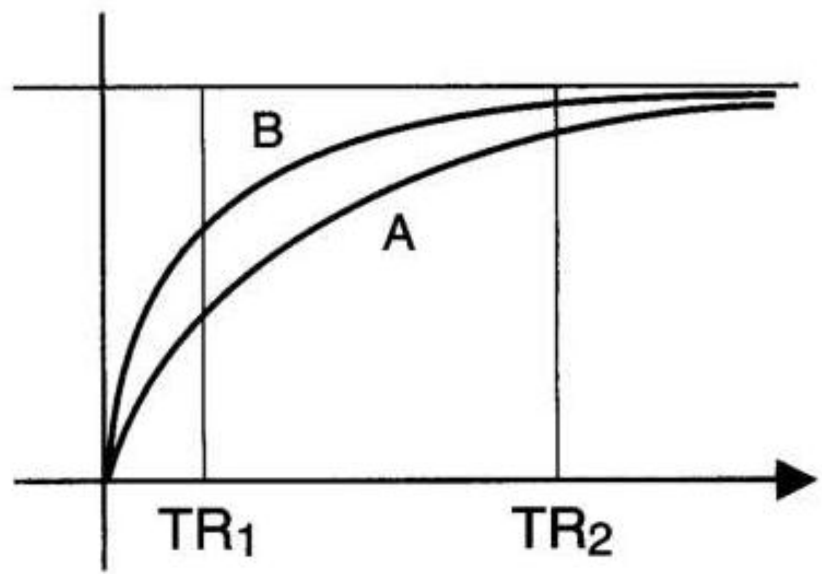


Signal Intensity = SI $\propto$ N(H)($e^{-TE/T2*}$) (1 - $e^{-TR/T1}$)

T1-Weighting

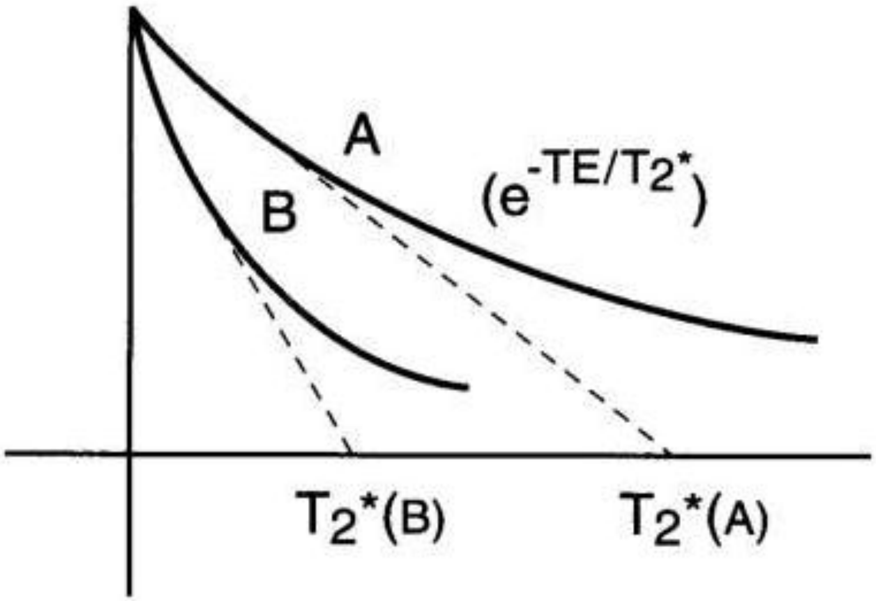
- Long TR reduces the T1 effect
- Short TR enhances the T1 contrast

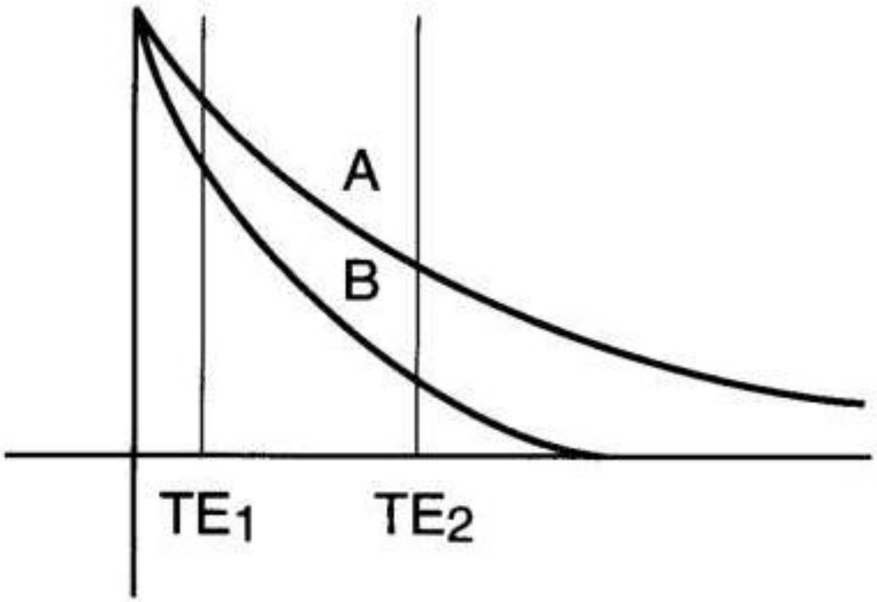




T2-Weighting

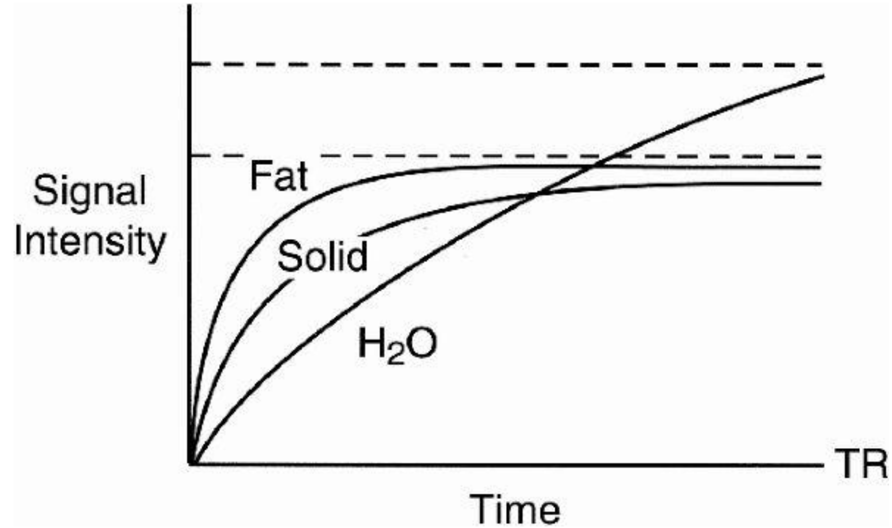
- Short TE reduces the T2* (T2) effect
- Long TE enhances the T2* (T2) effect

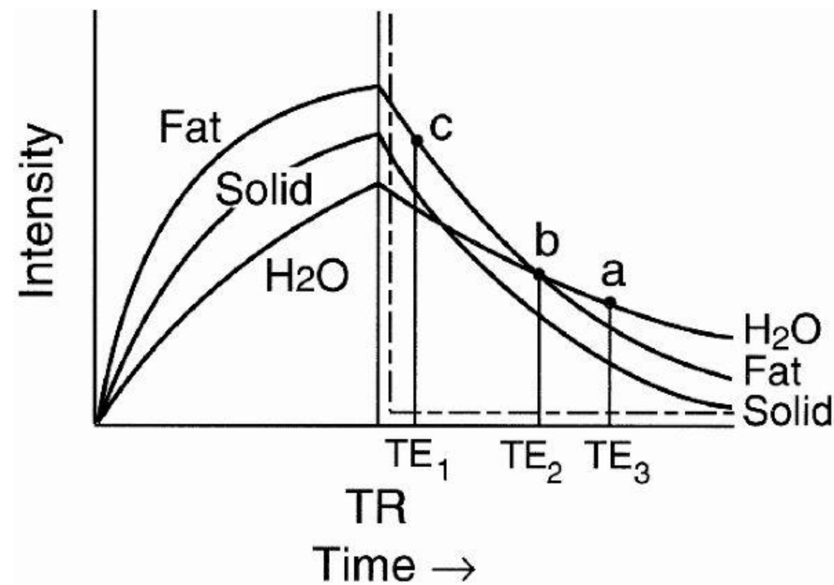




Tissue Contrast: Clinical Applications

- T1 Recovery Curve
  - Fat has the shortest T1
  - Proteinaceous fluid also has a short T1
  - H_2O has the longest T1
  - Solid tissue has intermediate T1
- T2 decay Curve
  - H_2O has a very long T2
  - Solid tissue has short T2
  - Fat has an intermediate T2
  - Proteinaceous fluid may have a short or intermediate T2 depending on the protein content





Summary of T1/T2 Values for Tissues

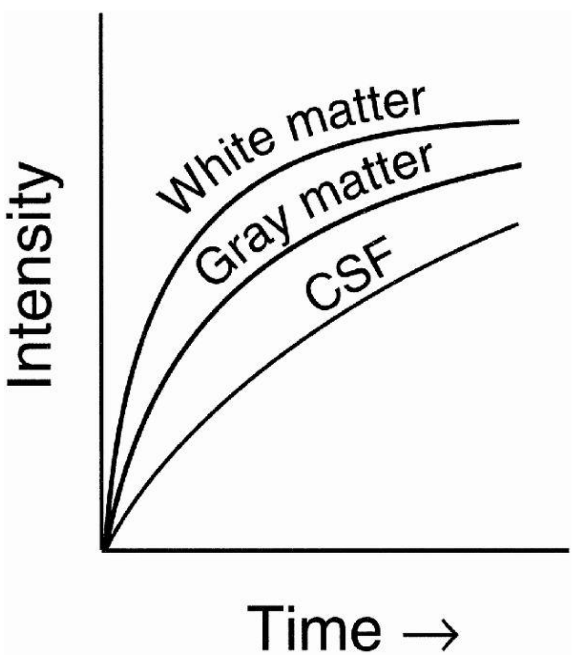
	Long T1 (low SI)	**Intermediate**	**Short T1 (high SI)**
Long T2 (high SI)	Water/CSF Pathology Edema		d (EC metHgb)
Intermediate		Muscle GM a (oxyHgb) WM	
Short T2 (low SI)	Air Cortical bone Heavy Ca^{++} b (deoxyHgb) e (hemosiderin) Fibrosis Tendons		Fat Proteinaceous solutions c (IC met Hgb) Paramagnetic materials (Gd, etc.)

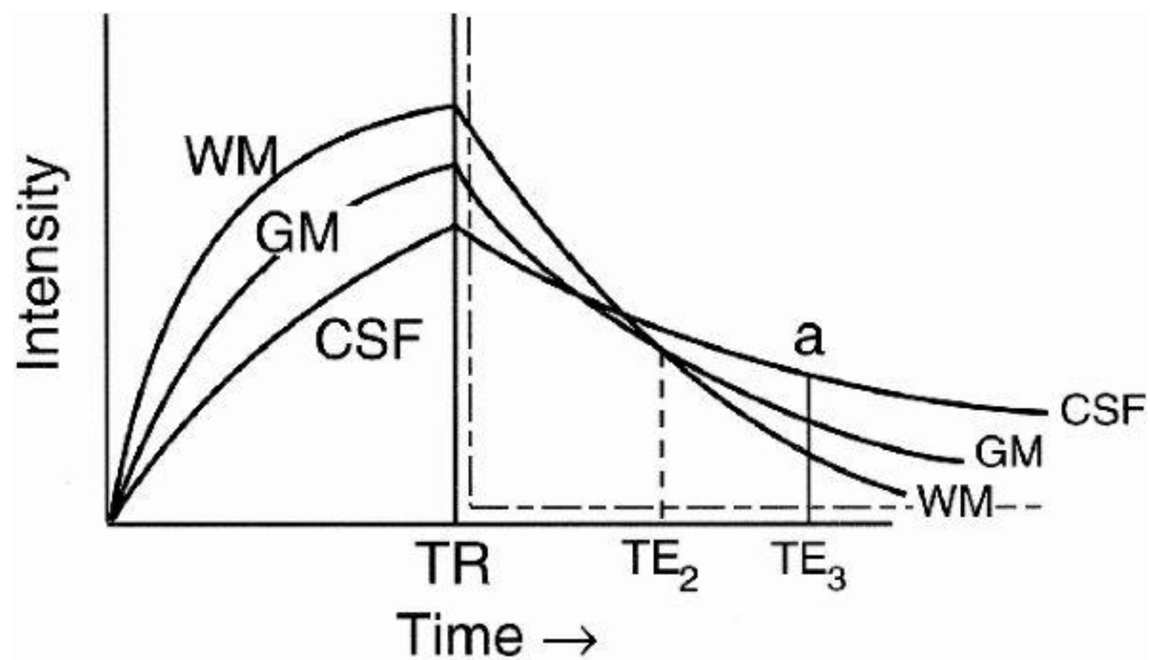
a-d represent breakdown products of hemoglobin (a, oxyhemoglobin; b, deoxyhemoglobin; c, intracellular methemoglobin; d, extracellular methemoglobin; e, hemosiderin). GM, gray matter; WM, white matter; SI, signal intensity; Hgb, hemoglobin; IC, intracellular; EC, extracellular.

Example: Brain Imaging

- WM: Fat
- GM: Solid Tissue
- CSF: H_2O

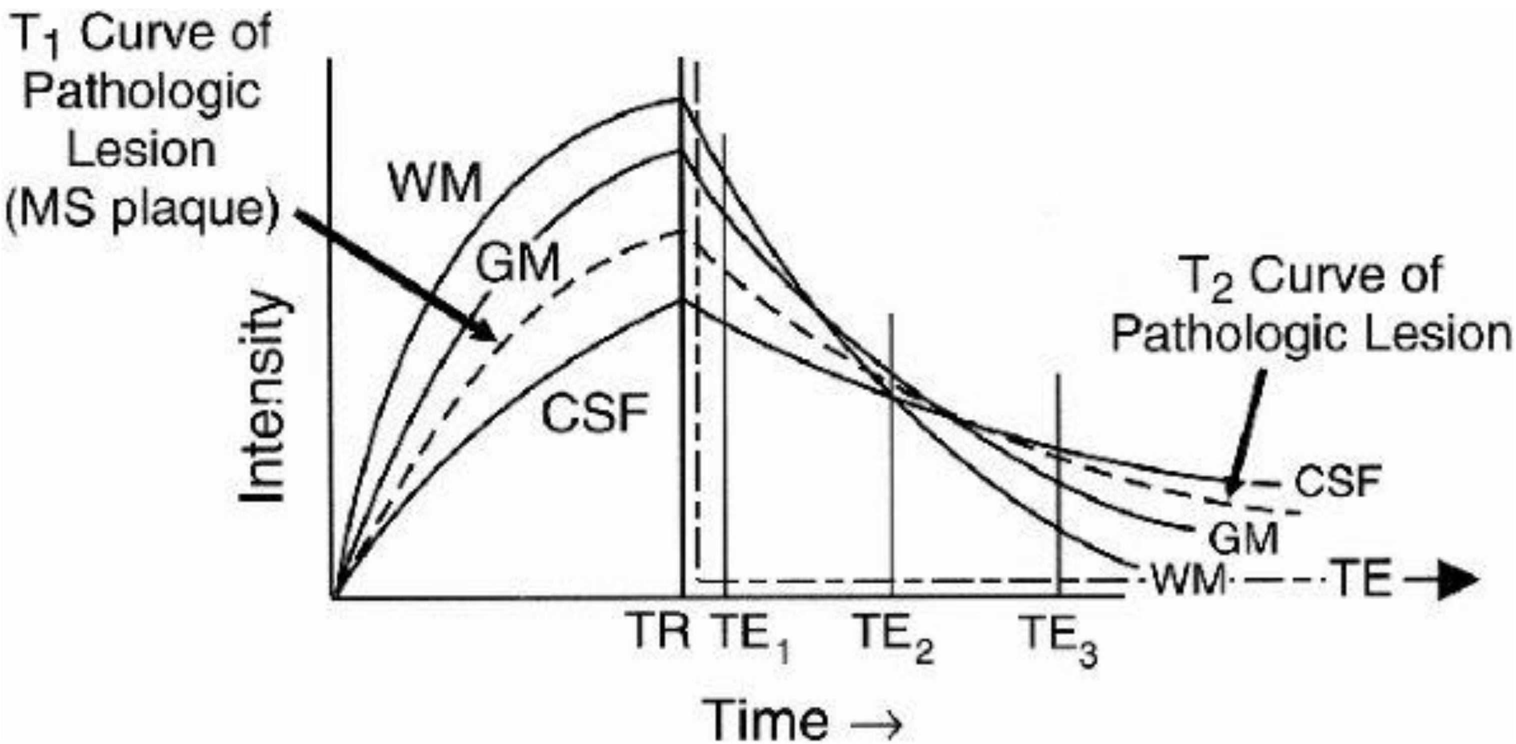
	T_1 (msec)	T_2 (msec)	N(H)
White matter	510	67	0.61
Gray matter	760	77	0.69
Edema	900	126	0.86
CSF	2650	180	1.00





Example: Brain Imaging

- Detecting a lesion
  - Compare contrasts at different TE values
  - TE1 appears to provide best contrast

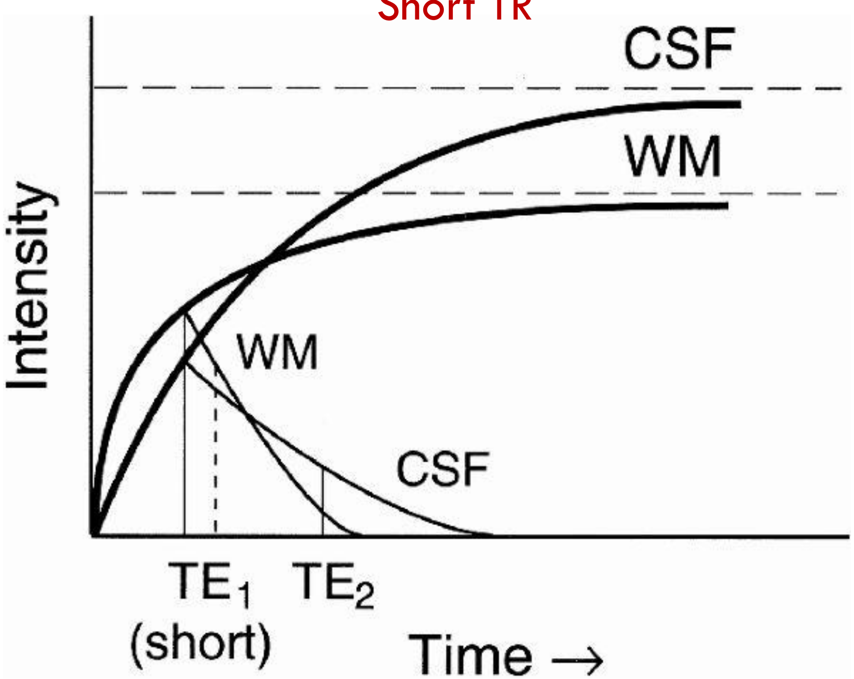


Tissue Contrast: T1W, T2W and PDW

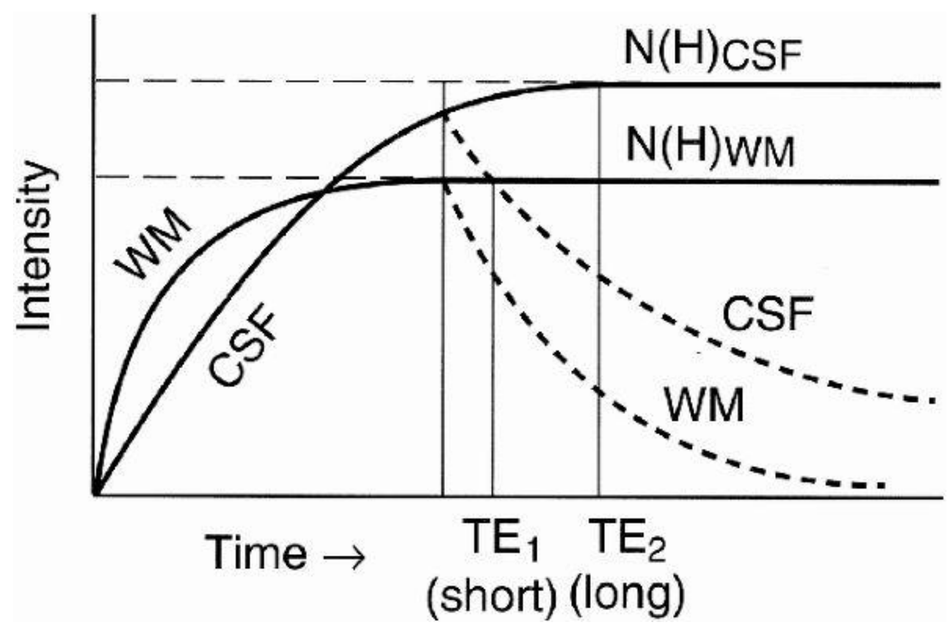
- Consider Cases of:
  - Short/Long TR
  - Short/Long TE

	T_1 (msec)	T_2 (msec)	N(H)
White matter	510	67	0.61
Gray matter	760	77	0.69
Edema	900	126	0.86
CSF	2650	180	1.00

Short TR



Long TR



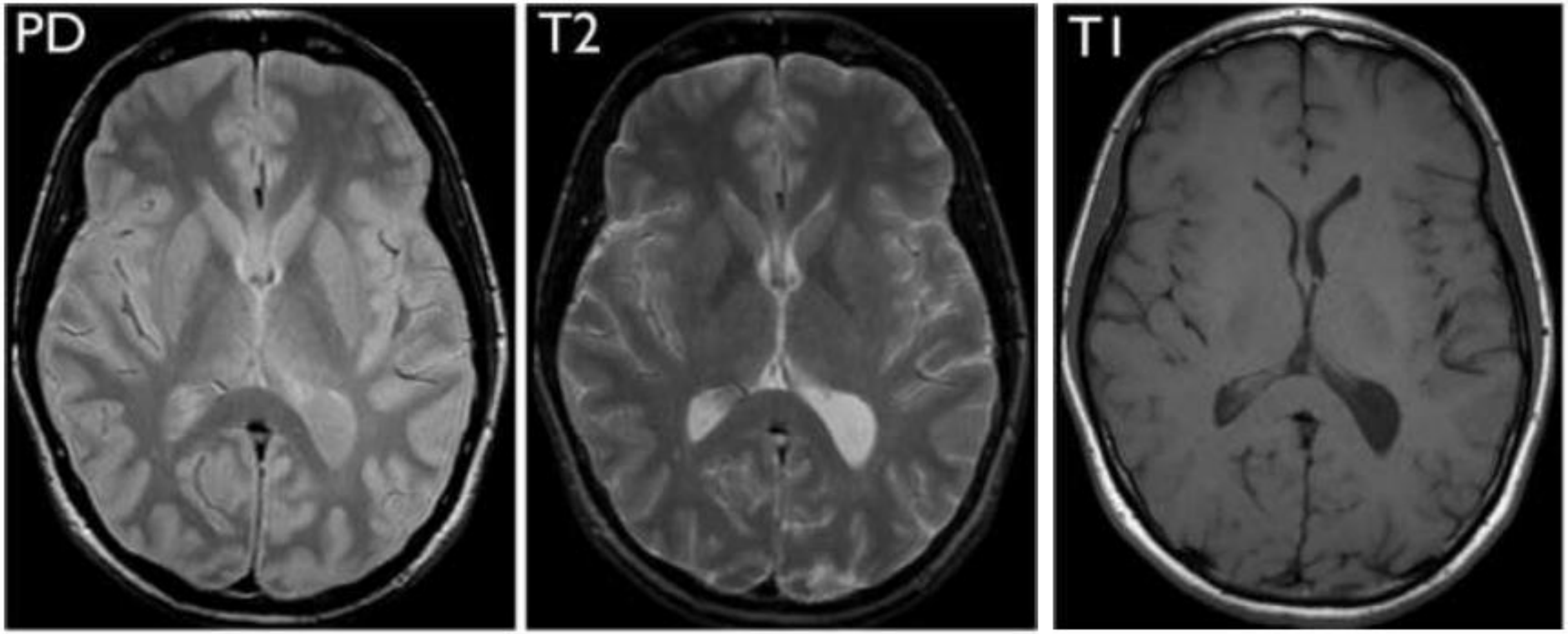
Tissue Contrast Summary

	TR	TE	Signal (Theoretical)
T1W	short	short	$N(H)(1 - e^{-TR/T1})$
T2W	long	long	$N(H)(e^{-TE/T2})$
PDW	long	short	$N(H)$

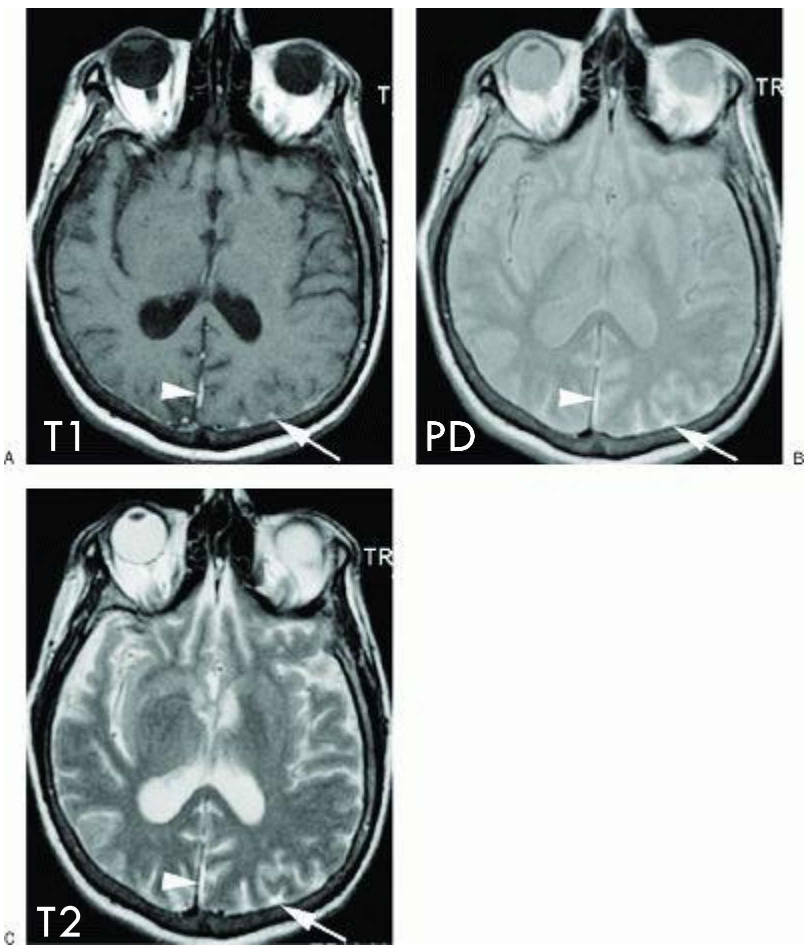
	Short TE	Long TE
short TR	T1W	mixed
long TR	PDW	T2W

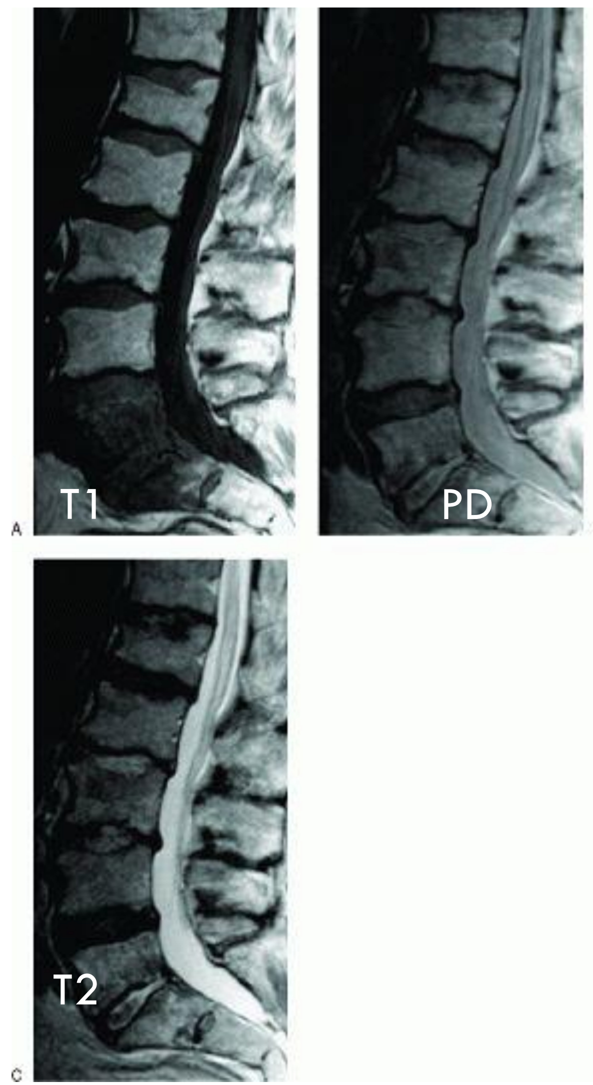
Tissue Contrast Examples

- Normal brain imaging
  - Very different contrast using different weighting selection

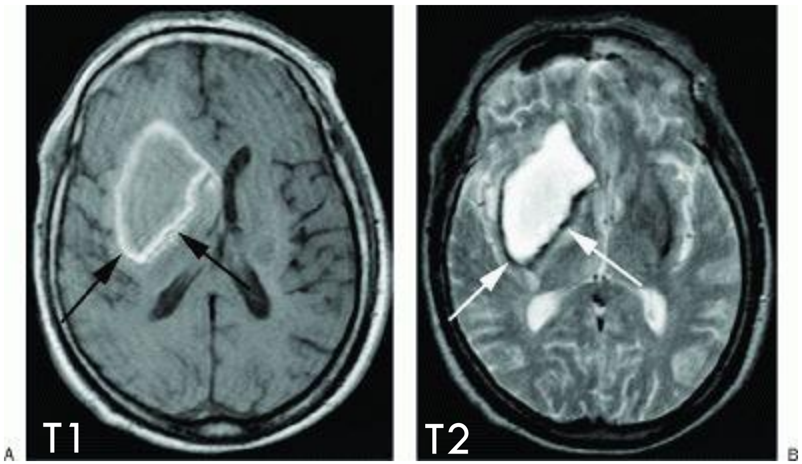


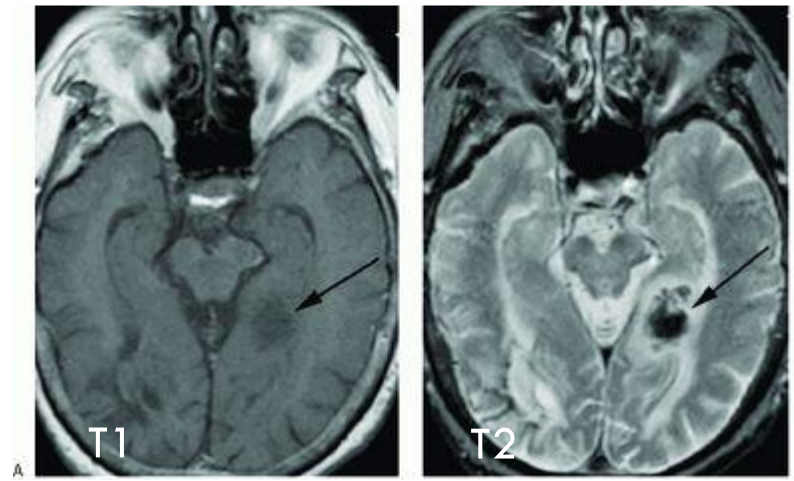
Tissue Contrast Examples

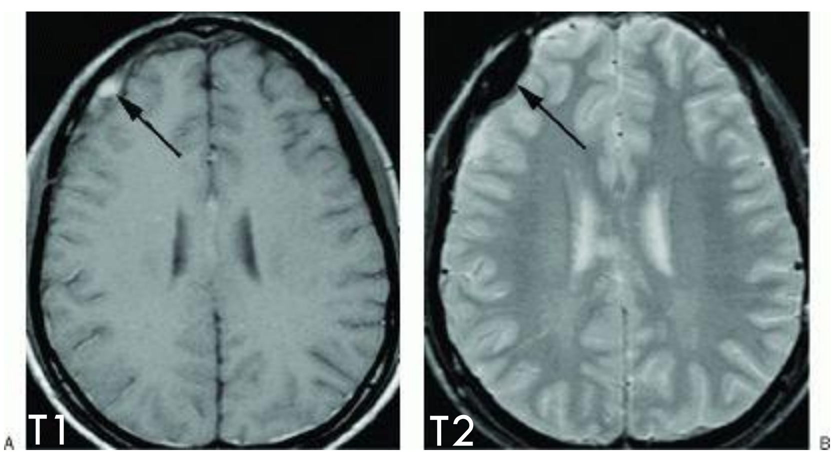




Tissue Contrast Examples

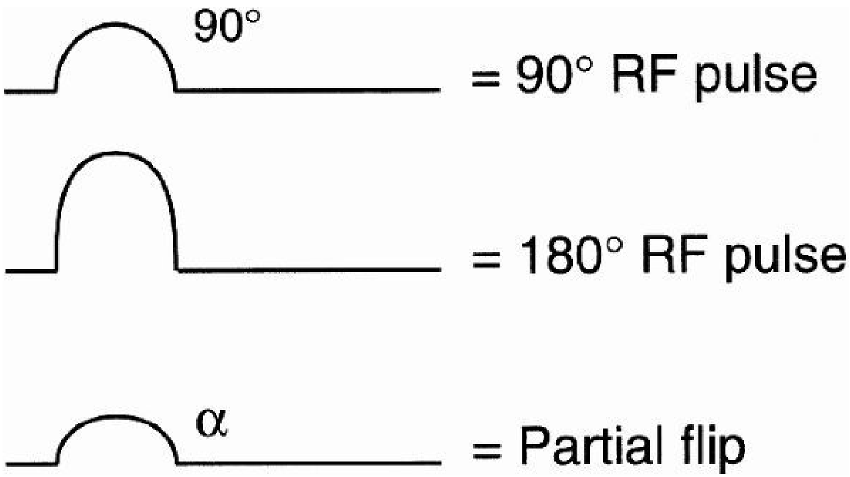






Pulse Sequences: Saturation, Saturation Recovery and Inversion Recovery

- A pulse sequence is a sequence of radio frequency (RF) pulses applied repeatedly during an MR study
  - Embedded in it are the TR and TE time parameters
- It is related to a timing diagram or a pulse sequence diagram
- 90° pulse: Saturation
- 180° pulse: Inversion
- <90° pulse: Partial Saturation

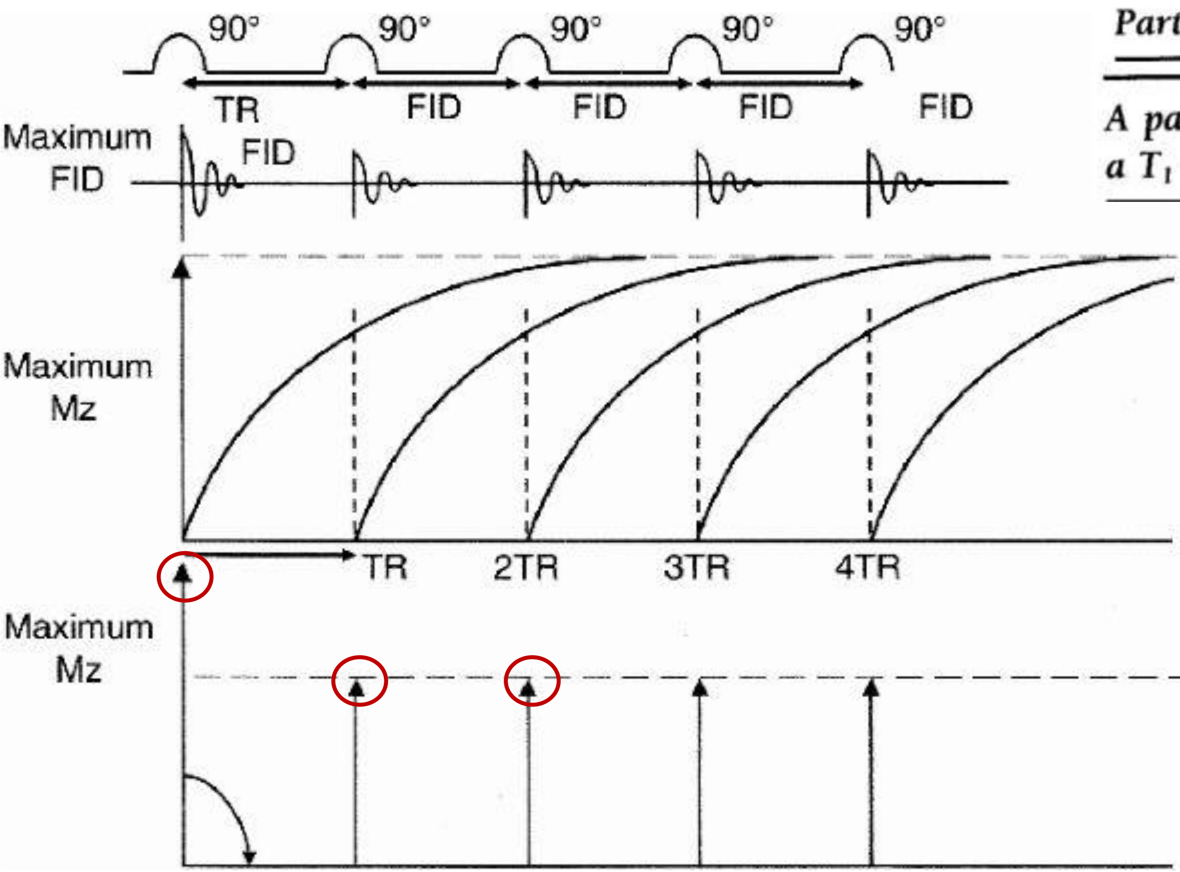


Saturation

- Immediately after the longitudinal magnetization has been flipped into the x-y plane by a 90° pulse, the system is said to be saturated
  - Application of a second 90° pulse at this moment will elicit no signal (like beating a dead horse).
- A few moments later, after some T1 recovery, the system is **partially saturated**
- With complete T1 recovery to the plateau value, the system is **unsaturated** or **fully magnetized**
- If longitudinal magnetization only partially flipped into the x-y plane (i.e., flip angles less than 90°), then there is still a component of magnetization along the z axis
  - Spins in this state are also **partially saturated**

Partial Saturation Pulse Sequence

- Start with a 90° pulse, wait for a short period TR, and then apply another 90° pulse. Keep repeating this sequence.



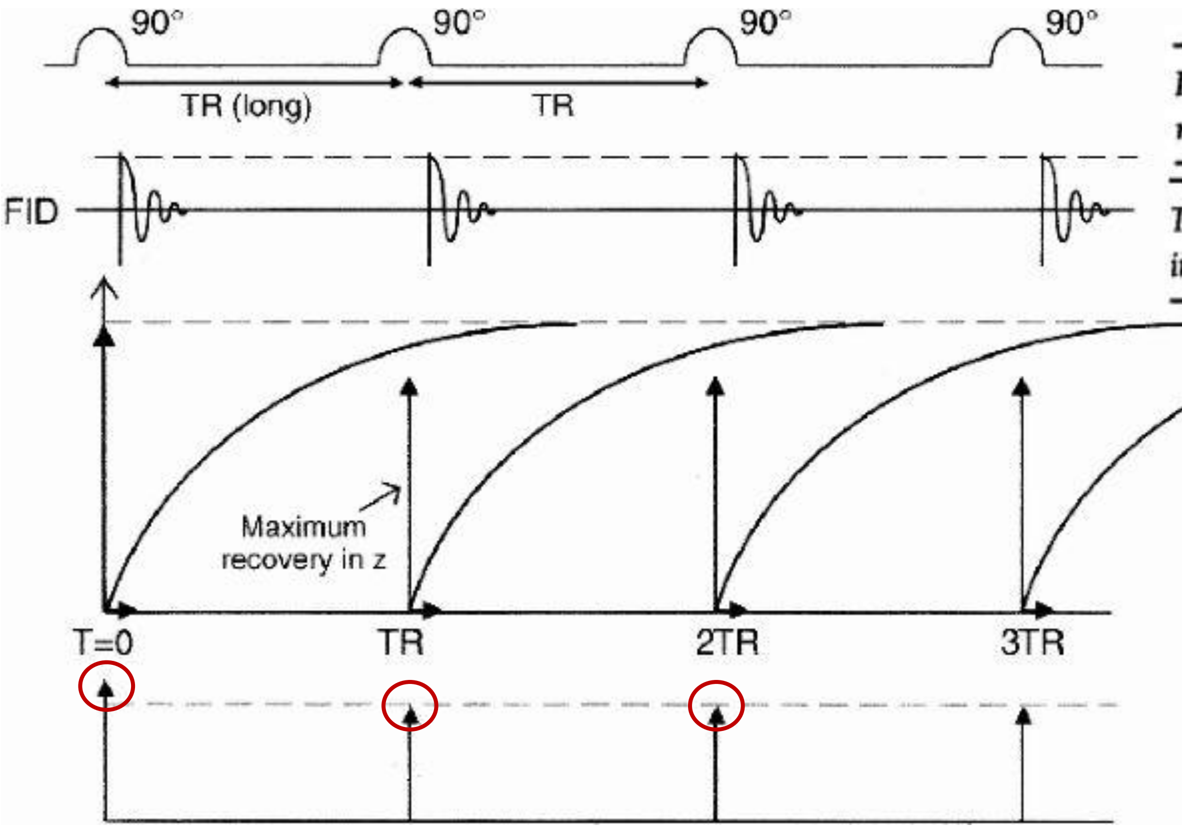
Partial saturation: TR is short, TE is minimal.

A partial saturation pulse sequence generates a T_1 weighted image.

Longitudinal Magnetization only partially recovered

Saturation Recovery Pulse Sequence

- We try to recover all the longitudinal magnetization before we apply another 90° RF pulse
  - Wait a long time before we apply a second RF pulse (Long TR)



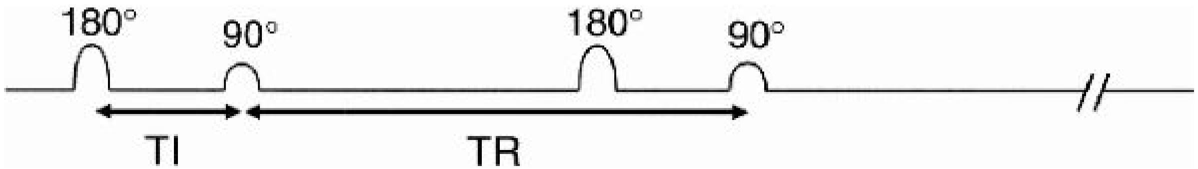
In saturation recovery, TR is long and TE is minimal.

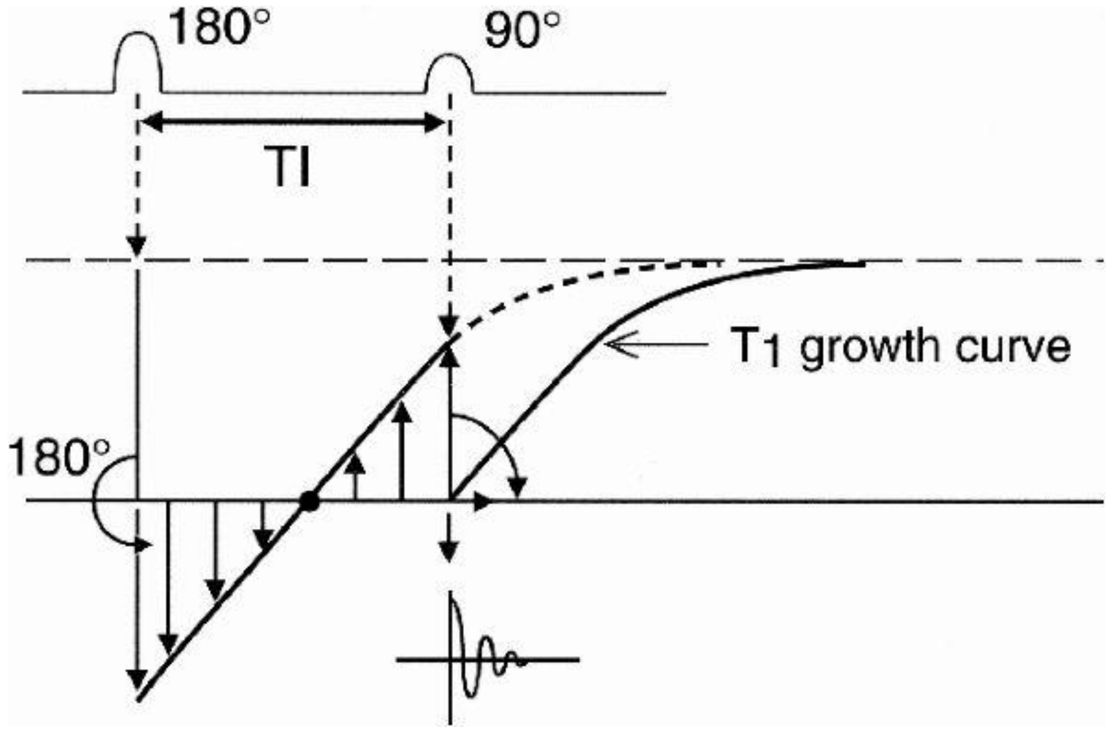
The saturation recovery pulse sequence results in a proton density weighted image.

Longitudinal Magnetization almost completely recovered

Inversion Recovery Pulse Sequence

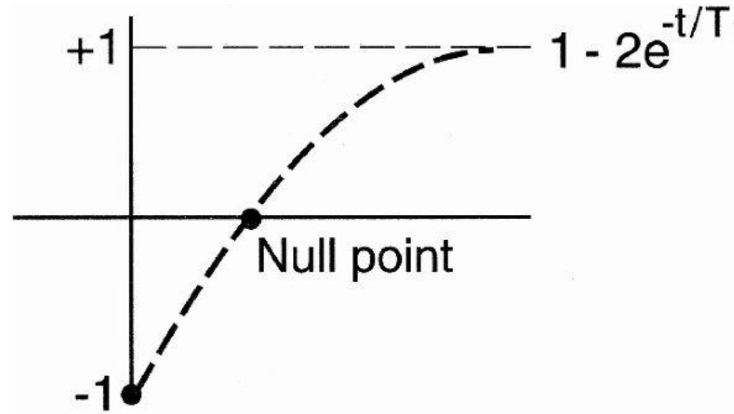
- first apply a 180° RF pulse. Next, we wait a period of time (the inversion time TI) and apply a 90° RF pulse

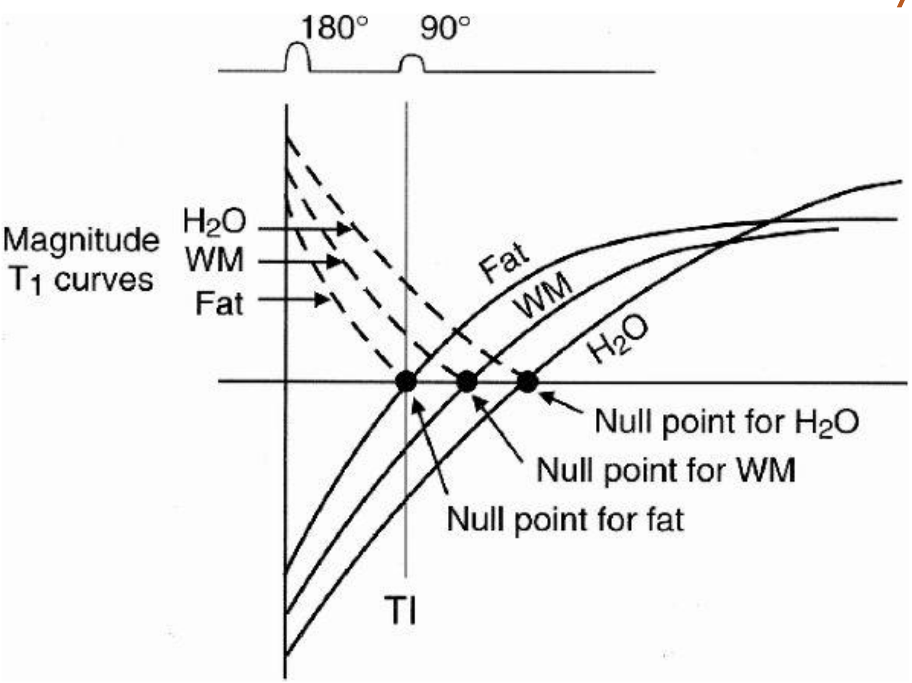




Inversion Recovery: Null Point

- The point at which the signal crosses the zero line is called the null point
  - Clinical application: Suppress a tissue
- Example: Fat Suppression using STIR
  - STIR: **S**hort **TI** **I**nversion **R**ecovery





$$\text{Signal intensity} = 0 = 1 - 2e^{-TI/T1}$$

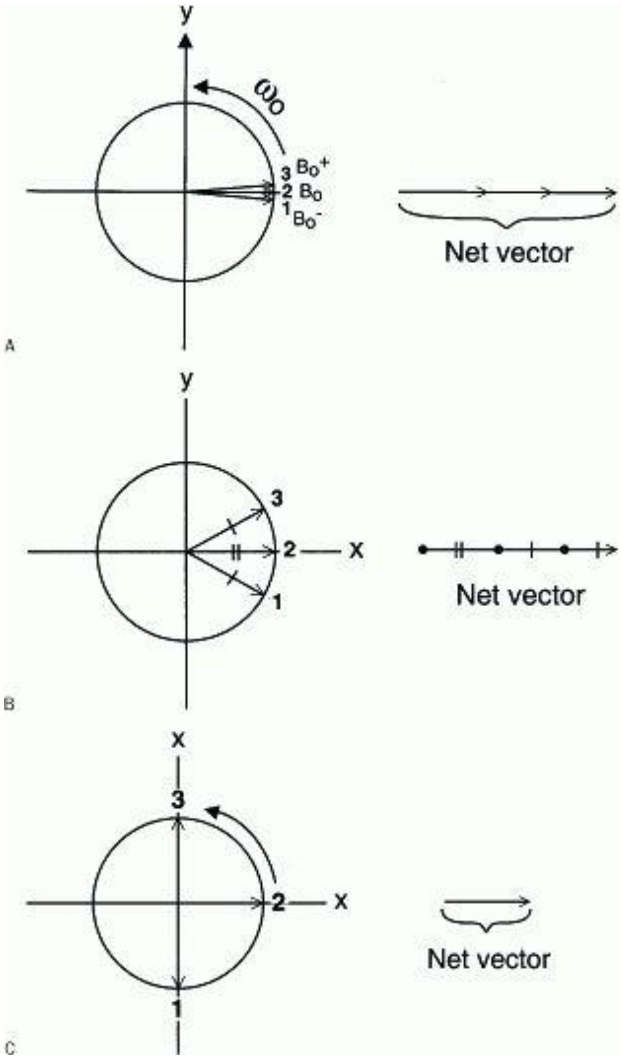
$$\boxed{TI\ (\text{null}) = 0.693 \times T1}$$

Pulse Sequences: Spin Echo

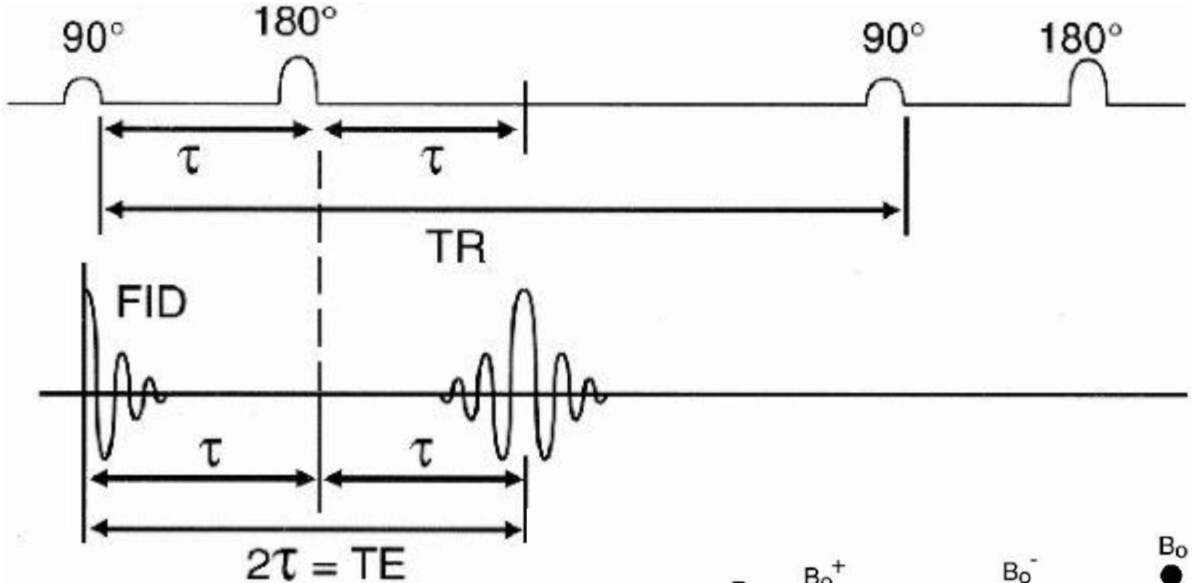
- Dephasing problem causes
  - Spin-spin interactions (inherent)
  - External magnetic field inhomogeneity

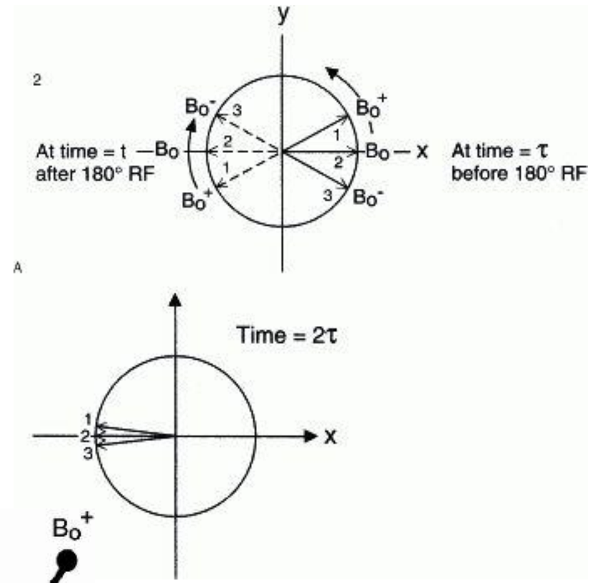
$$\boxed{1/T2^{*} = 1/T2 + \gamma\Delta B}$$

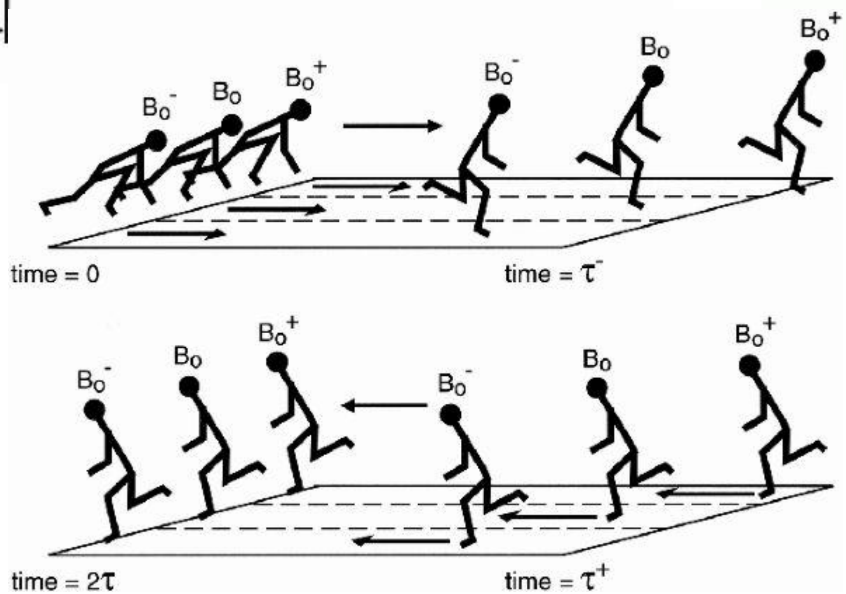
- Spin echo sequence: only T2



Spin Echo Pulse Sequence

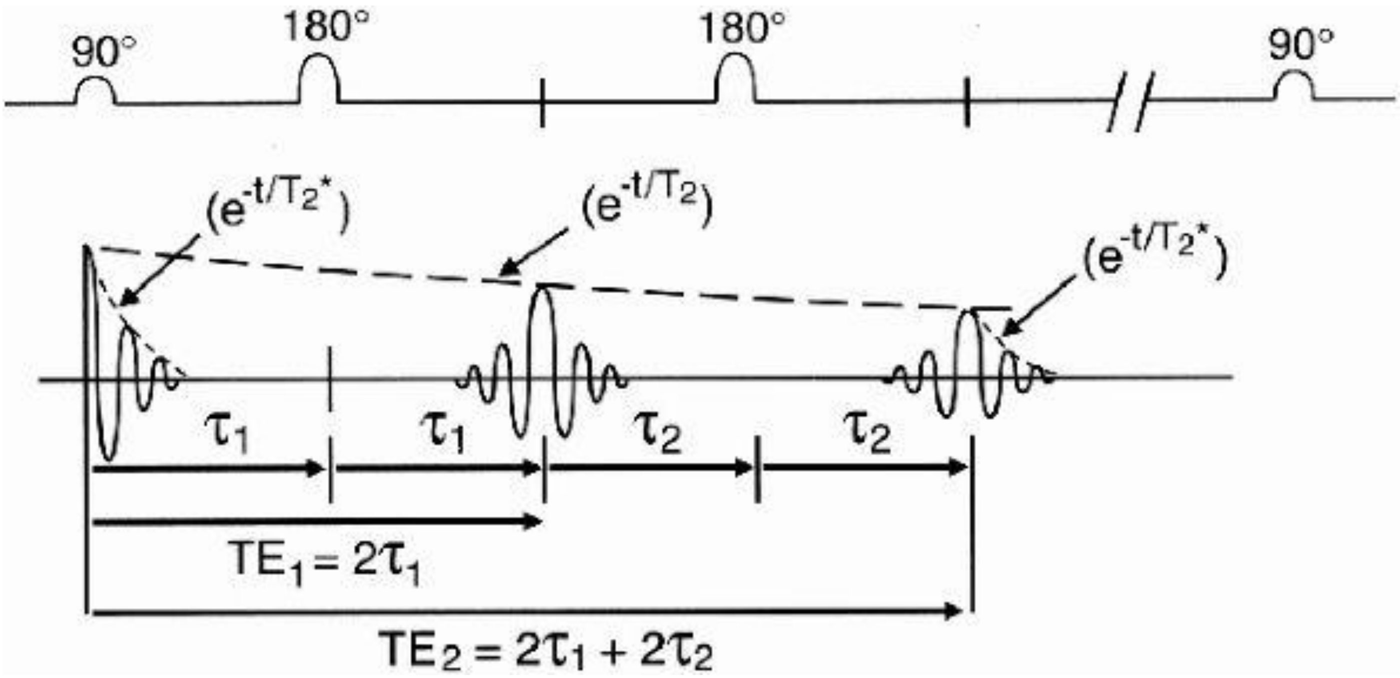






Multi-Echo Spin Echo Pulse Sequence

- Add another 180° rephasing pulse
  - Symmetric echoes: $\tau_1 = \tau_2$
  - Asymmetric echoes: $\tau_1 \neq \tau_2$



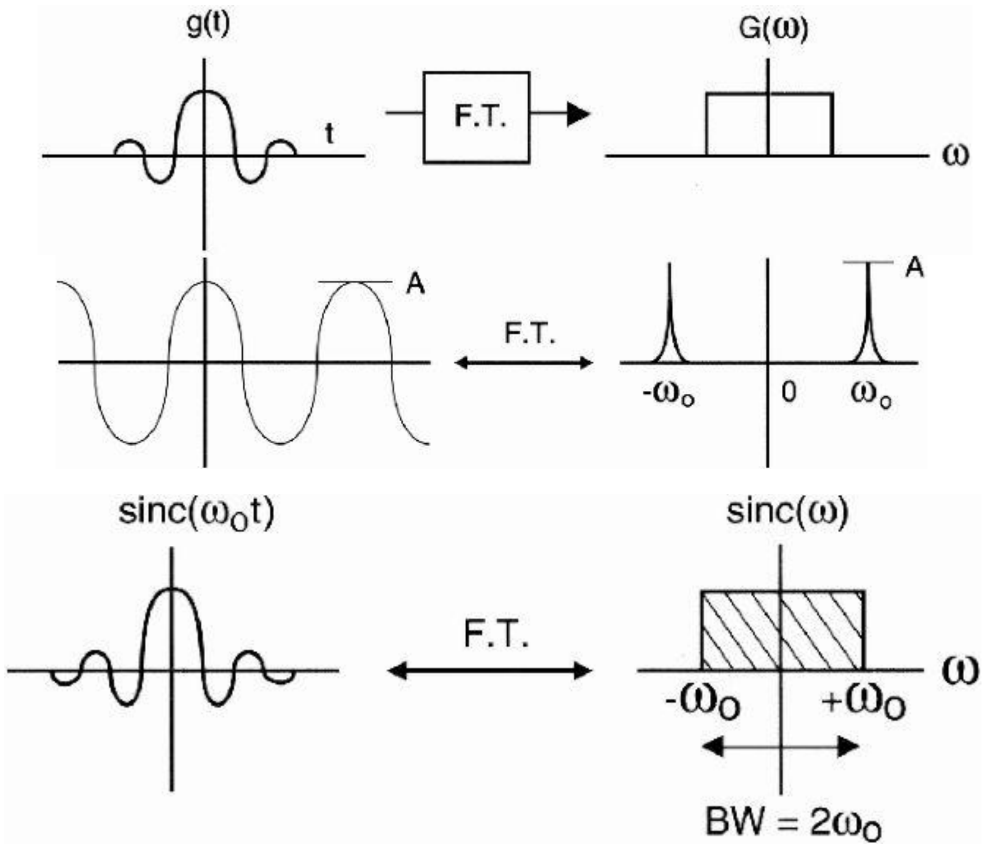
Tissue Contrast with Spin Echo

Contrast	TR	TE	Signal (Theoretical)
T1W	Short	Short	$N(H)(1-e^{-TR/T1})$
T2W	Long	Long	$N(H)(e^{-TE/T2})$
PDW	Long	Short	$N(H)$

	Short TE	Long TE
Short TR	T1W	Mixed
Long TR	PDW	T2W

Fourier Transform

- The Fourier Transform (FT) provides a frequency spectrum of a signal.
  - It is sometimes easier to work in the frequency domain



Fourier Transform

- Forward transform (*Analysis*)

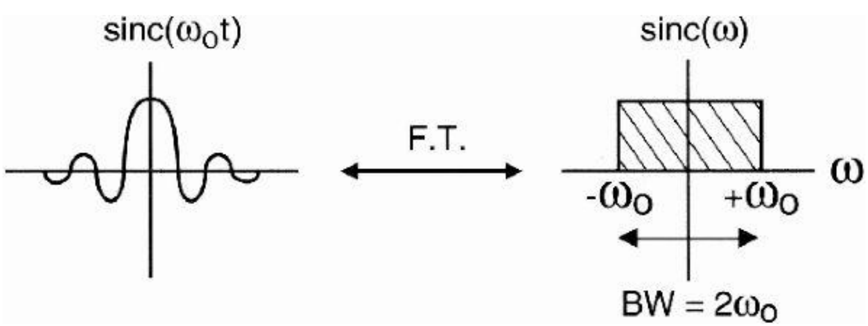
$$\mathcal{F}\{g\} = \iint_{-\infty}^{\infty} g(x, y) \exp\left[-j2\pi(f_X x + f_Y y)\right] dx\, dy.$$

- Inverse transform (*Synthesis*)

$$\mathcal{F}^{-1}\{G\} = \iint_{-\infty}^{\infty} G(f_X, f_Y) \exp[j2\pi(f_X x + f_Y y)]\, df_X\, df_Y.$$

Fourier Transform

- Effect of high frequencies
  - Details of signal
  - The more you acquire, the higher the resolution the image will be
  - The bandwidth (BW) is simply a measure of the range of frequencies present in the signal



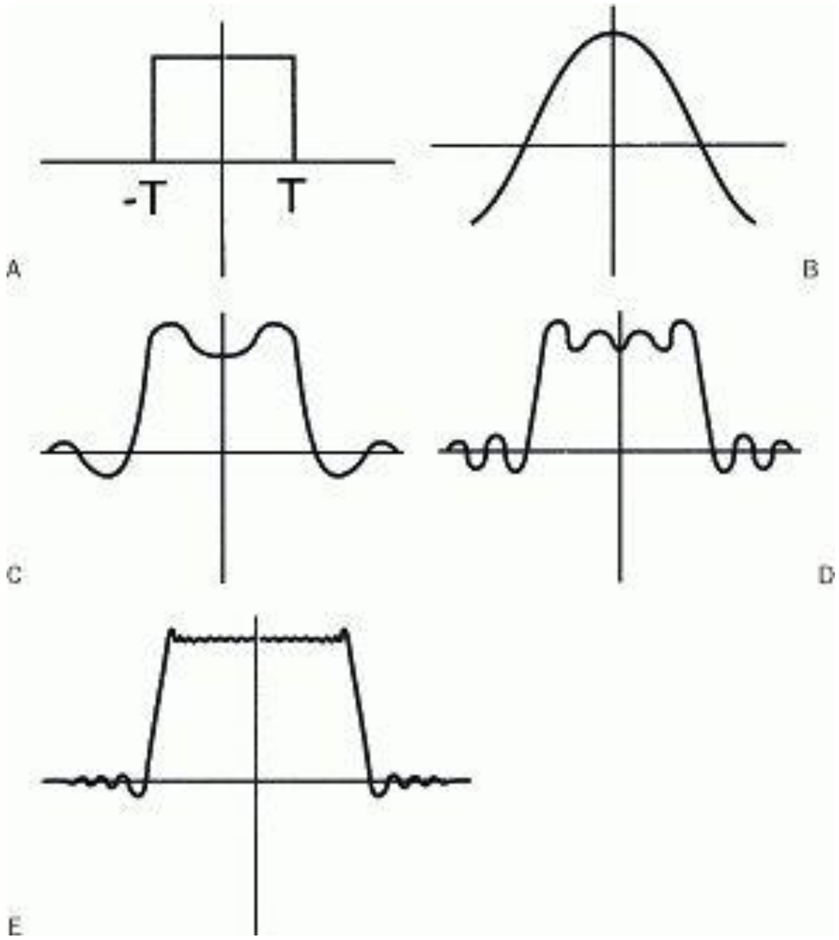
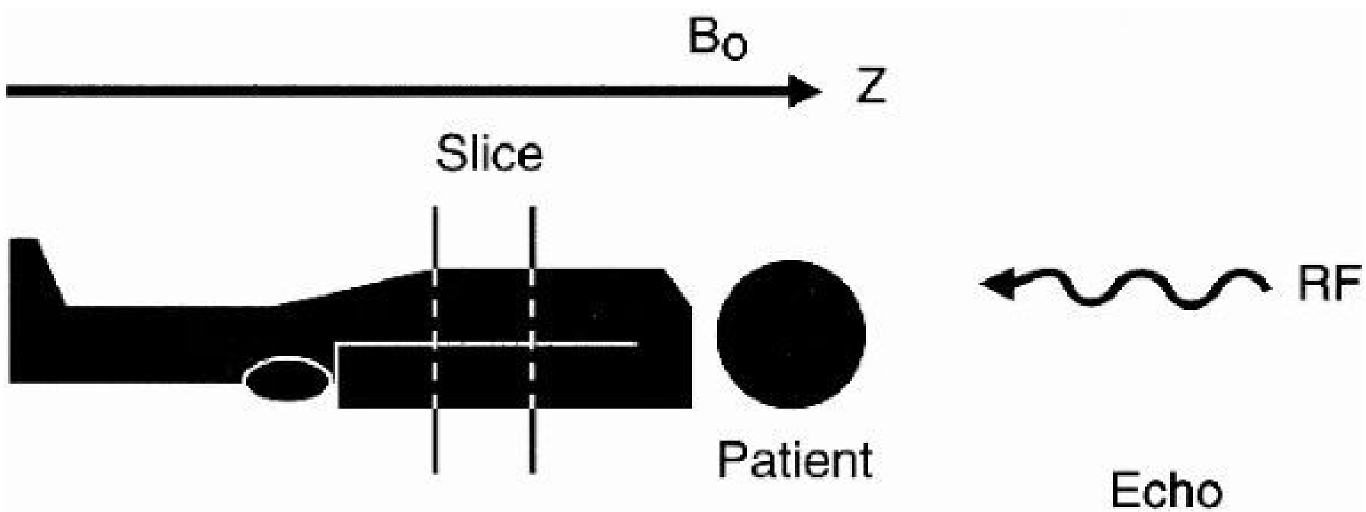


Image Reconstruction

- The signals received from a patient contain information about the entire part of the patient being imaged.
  - They do not have any particular spatial information. That is, we cannot determine the specific origin point of each component of the signal.
- This is the function of the gradients where one gradient is required in each of the x, y, and z directions to obtain spatial information in that direction.
  - Slice-select gradient
  - Readout or frequency-encoding gradient
  - Phase-encoding gradient
- Depending on their orientation axis they are called G_x, G_y, and G_z.
- Depending on the slice orientation (axial, sagittal, or coronal), G_x, G_y, and G_z can be used for slice select, readout, or phase encode.

Slice Selection

- Signal is obtained only from a particular slice from the body.
  - Can be in any direction



Slice Selection

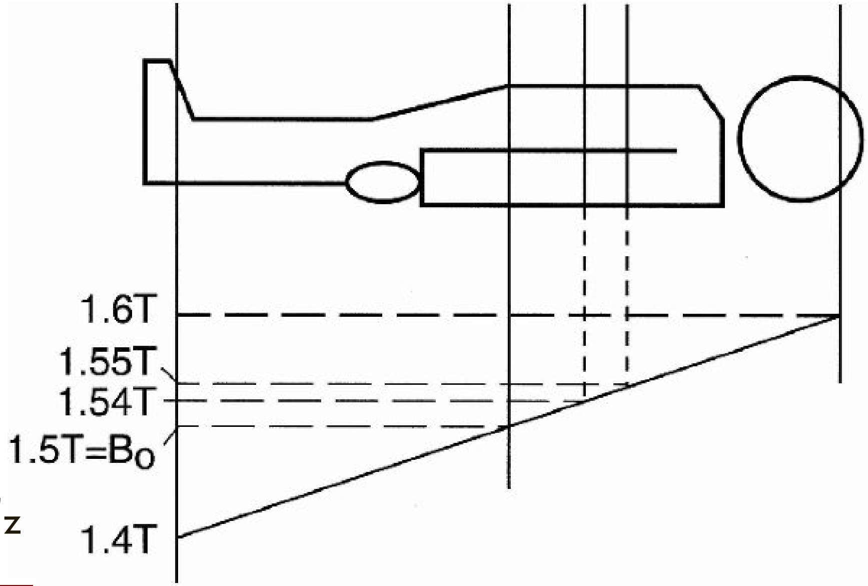
- Larmor equation:

$$\omega_o = \gamma B_0$$

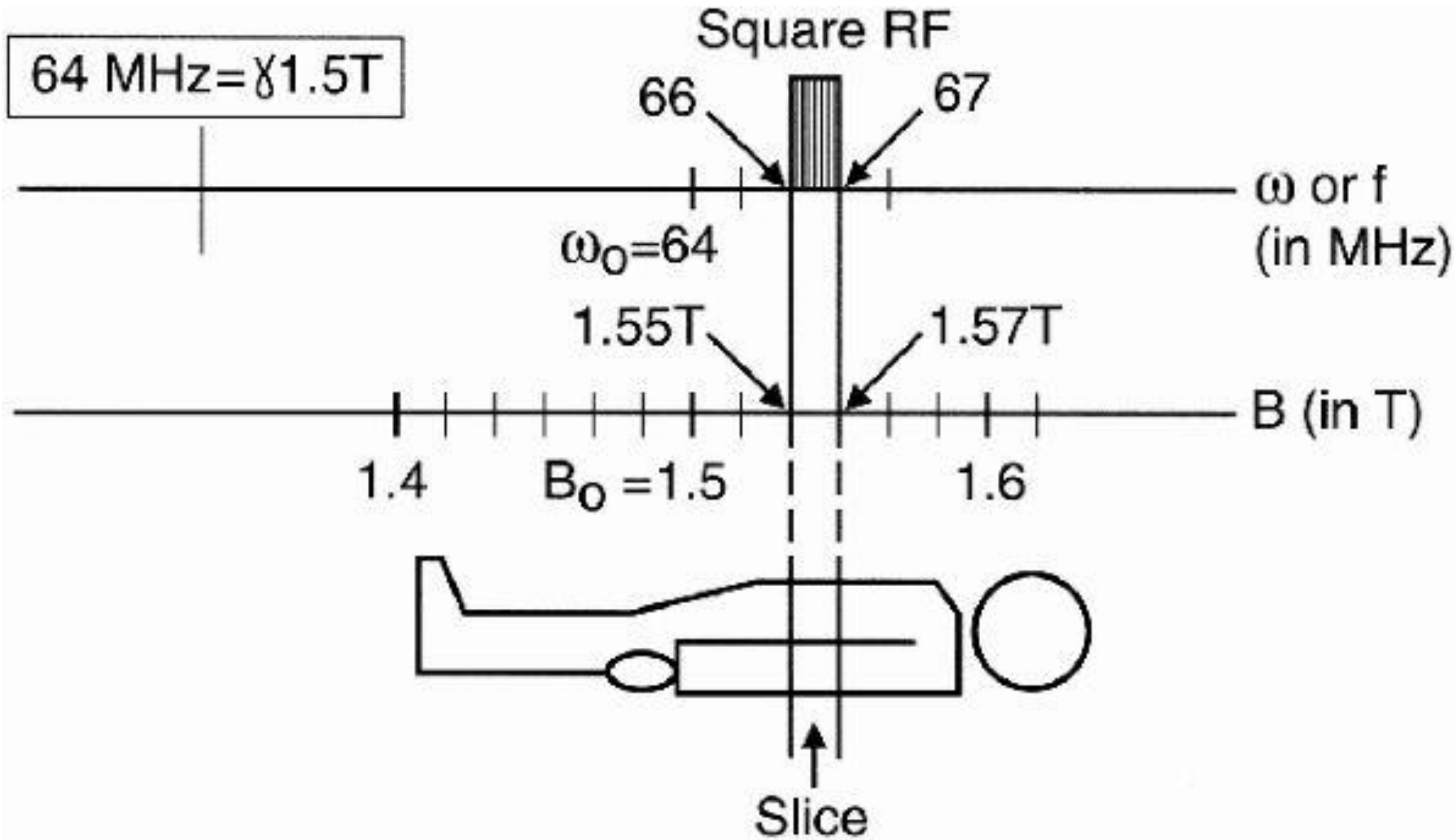
- Larmor equation with gradient G_z

$$\omega_o(z) = \gamma(B_0 + G_z \cdot z)$$

  - Larmor frequency depends on location
  - Send RF pulse with desired frequency range to excite a slice !

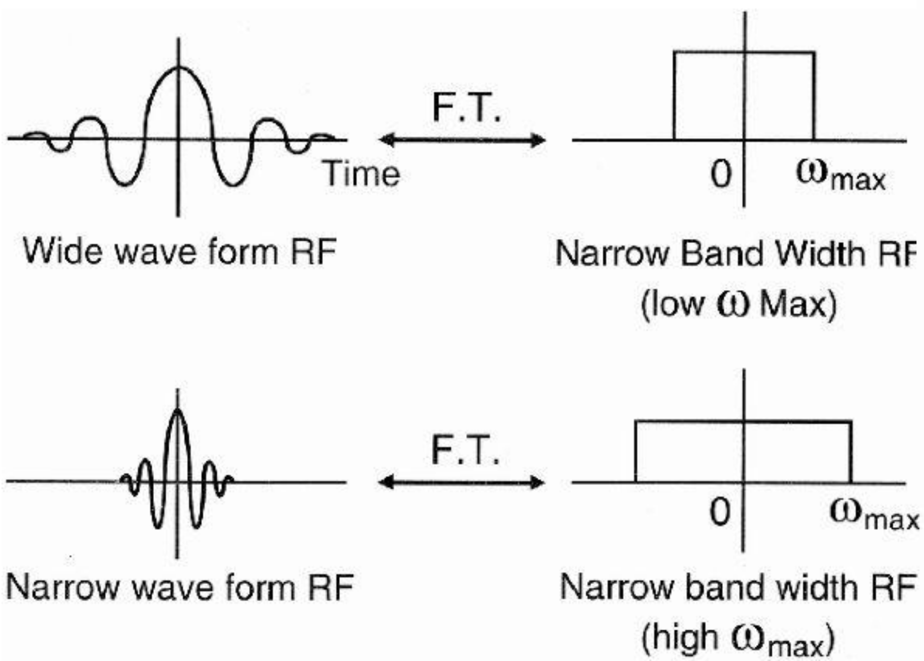


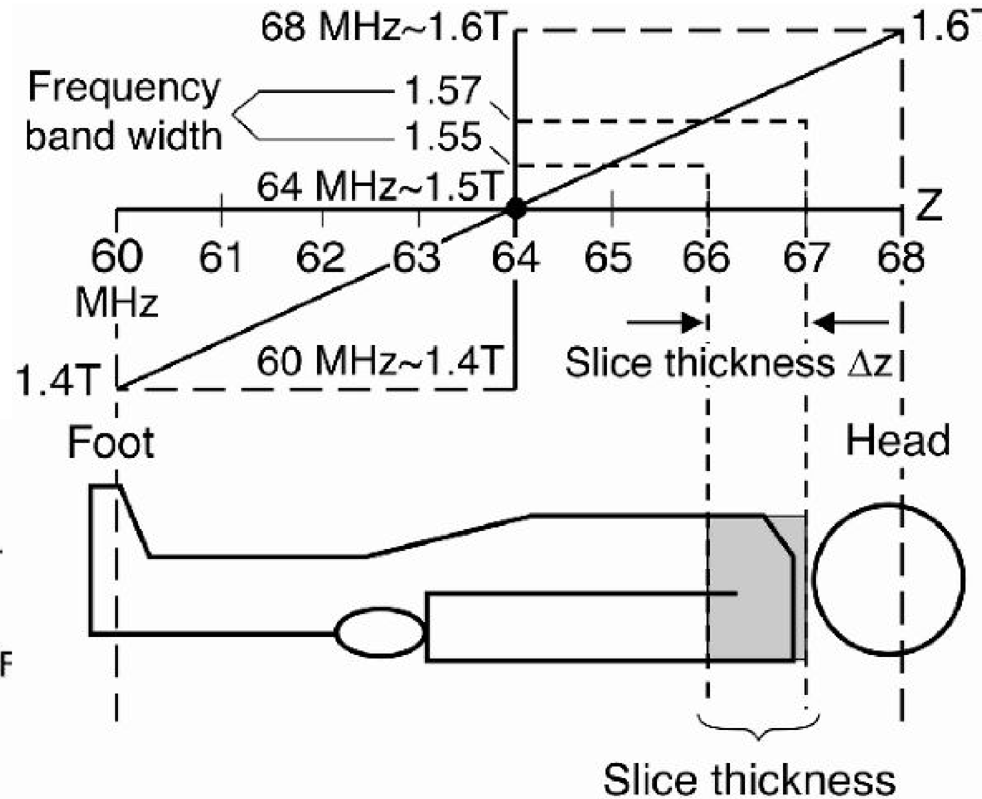
Slice Selection



Slice Selection

- Slice definition
  - Slice location
  - Slice thickness
  - Slice profile

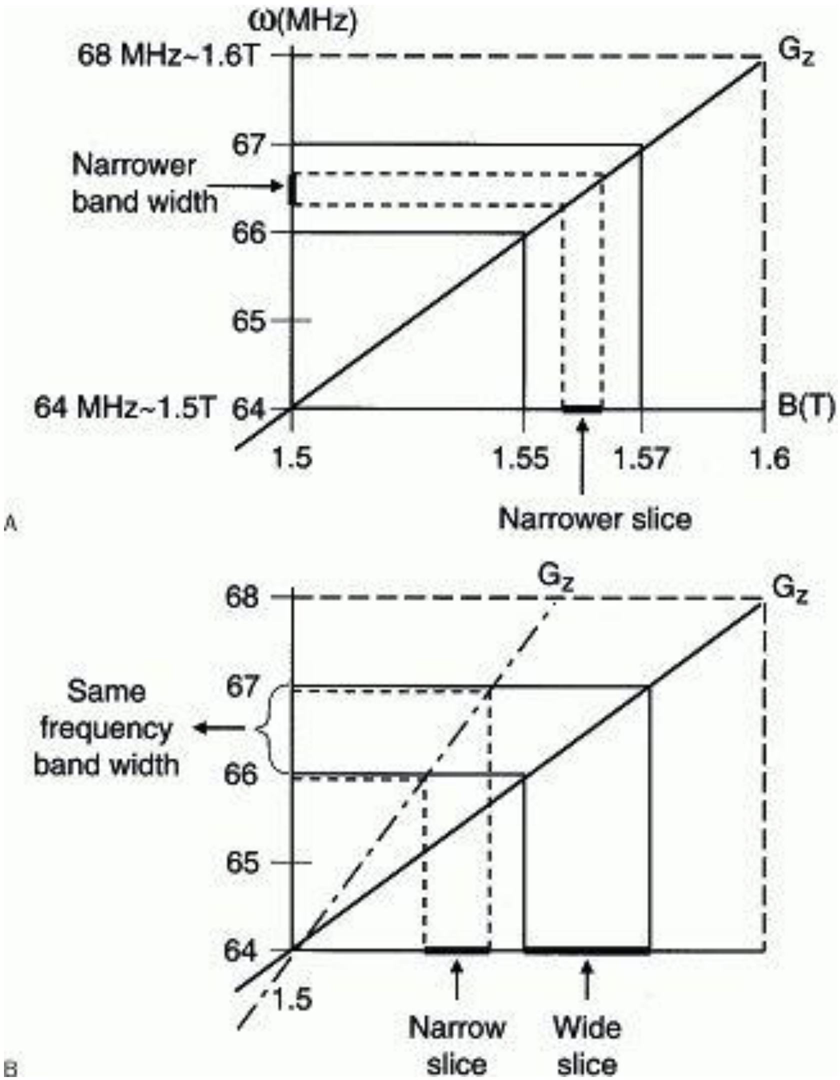




Slice Selection: Changing Thickness

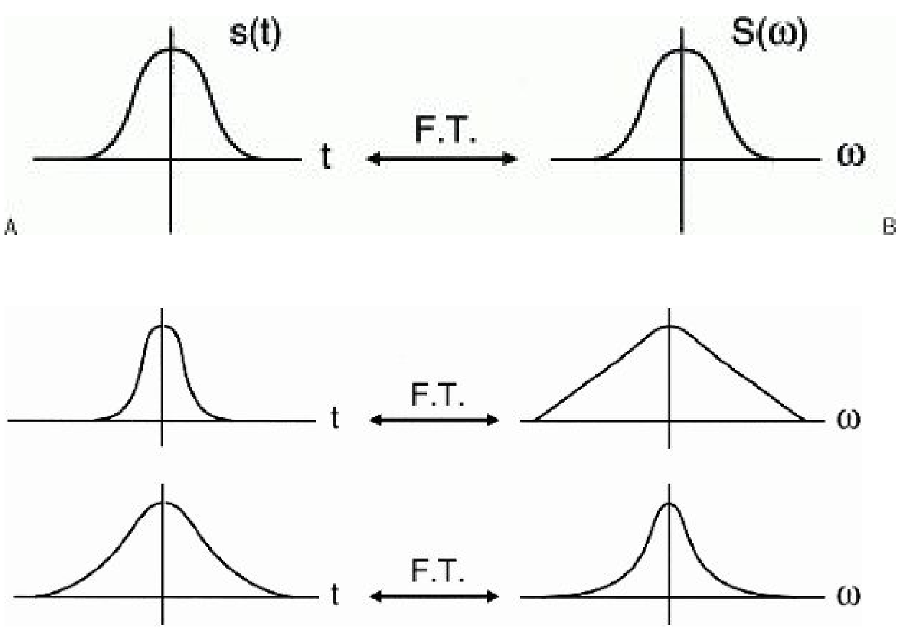
- Different RF pulse bandwidth
- Different slice selection gradient

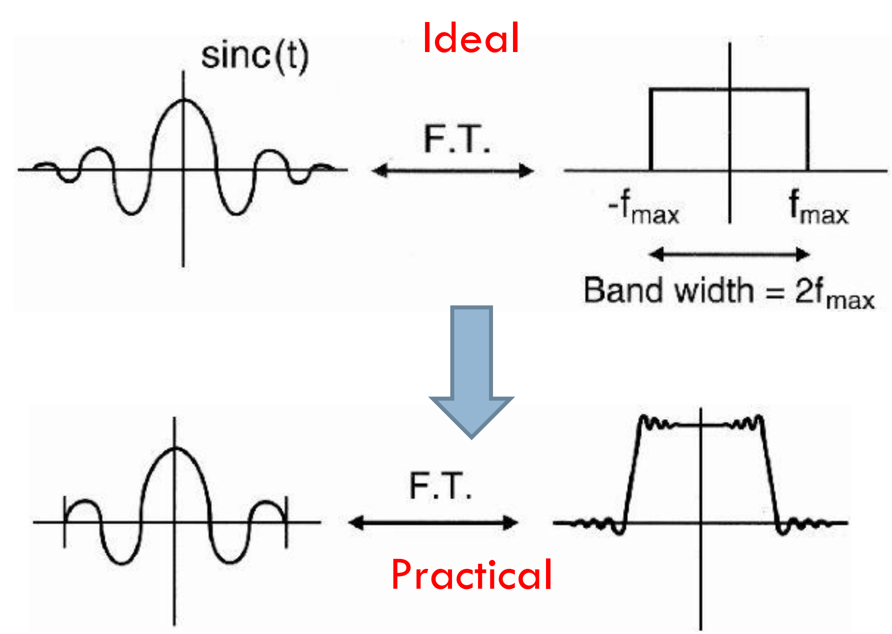
- To decrease the thickness is to use a narrower bandwidth.
  - Narrower frequency bandwidth will excite protons in a narrower band of magnetic field strengths
- Second way to decrease slice thickness is to increase the slope of the magnetic field gradient



Slice Selection: RF Pulses

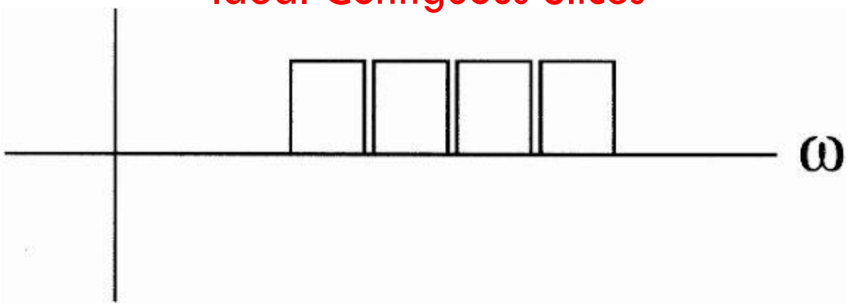
- There are two types of RF pulses:
  - Nonselective
  - Selective
- Slice profile = Fourier transform of pulse shape

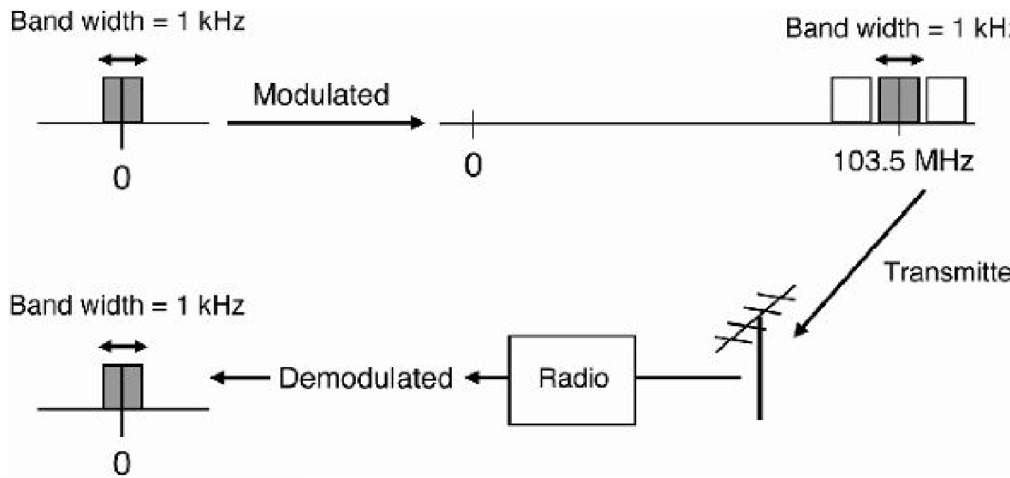




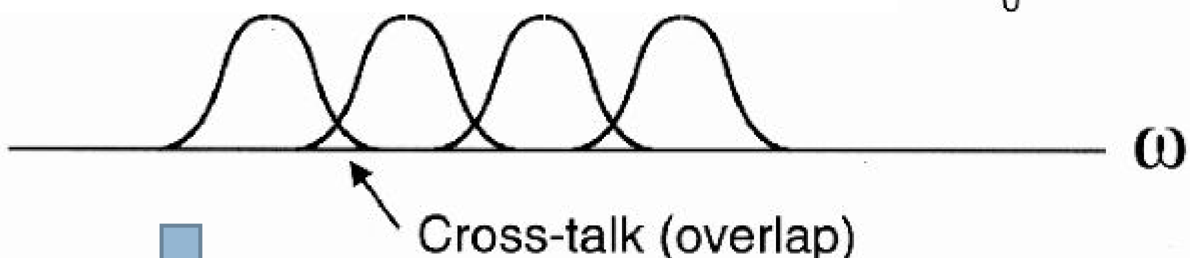
Slice Selection: Multi-Slice Scan

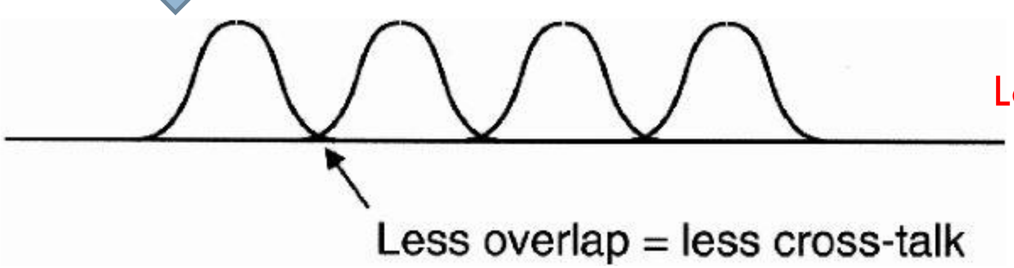
Ideal Contiguous Slices





Practical Slices





Leave more spacing between slices

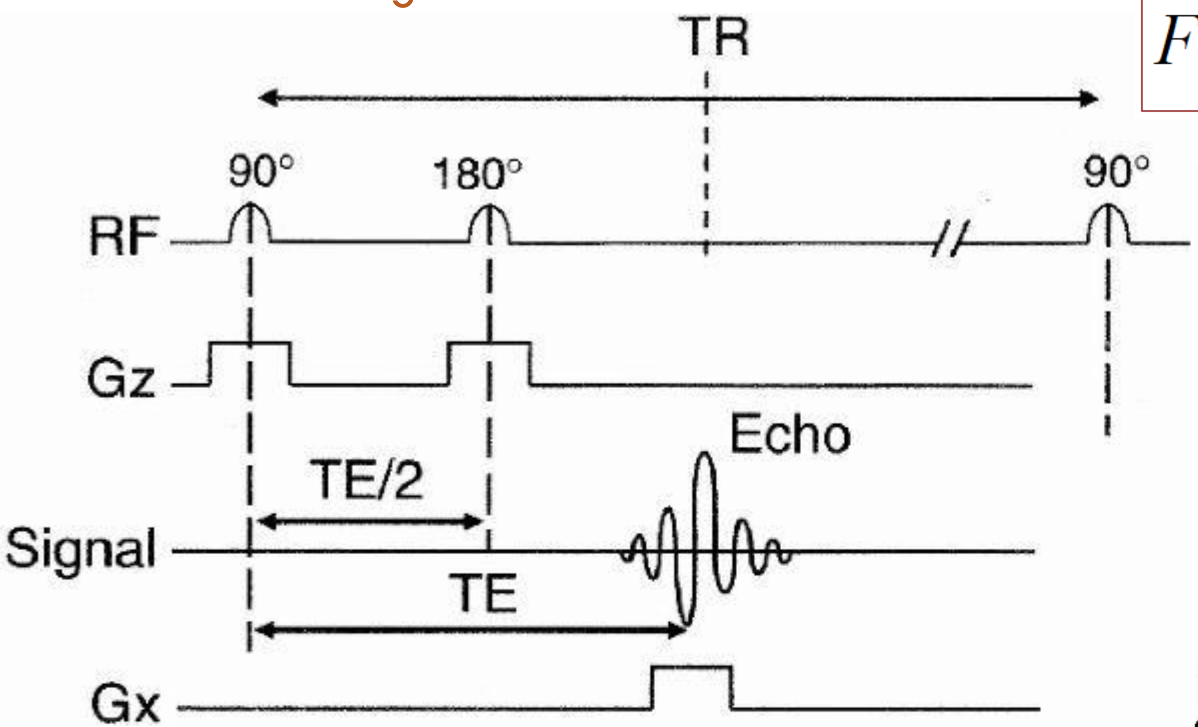
In-Plane Spatial Encoding: Fourier Imaging

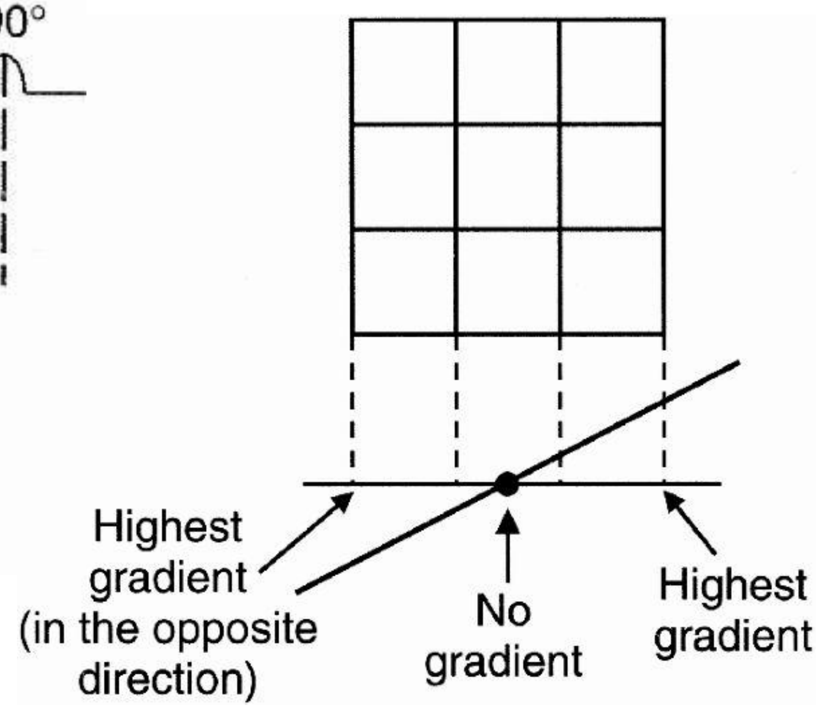
- Basic idea: encode location by frequency
  - Magnetic field gradient is used during reception
  - Larmor frequency depends on present magnetic field
  - Returned frequency from an area depends on its location
  - Easily decoded by Fourier transformation
- Applied by 2 different methods
  - Frequency encoding
  - Phase encoding

Frequency Encoding

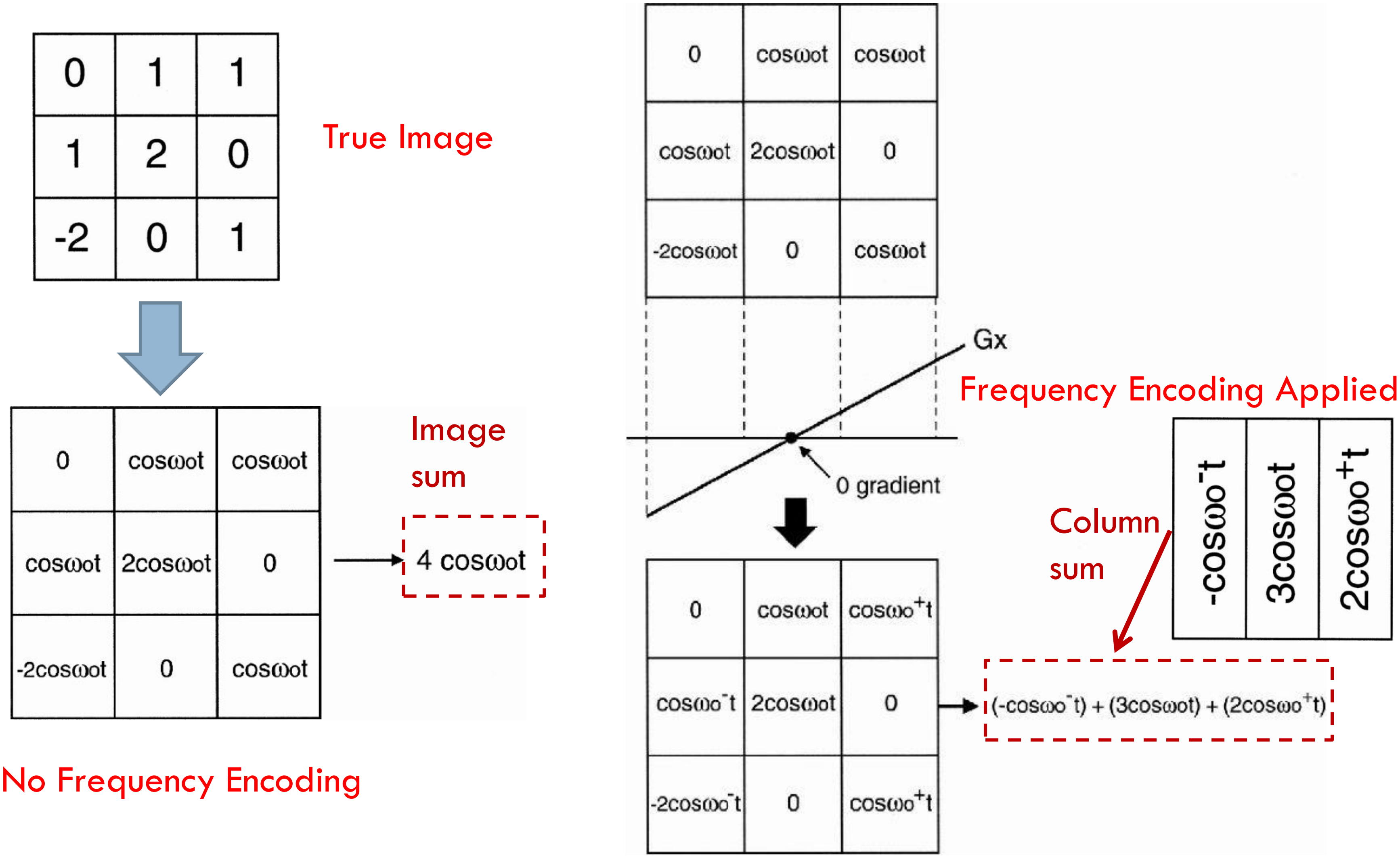
- Read-out gradient
  - The G_x gradient is applied during the time the echo is received, i.e., during readout

$$F(k_x) = \int f(x) \cdot e^{-j2\pi k_x \cdot x} dx$$





Frequency Encoding Example



Phase Encoding

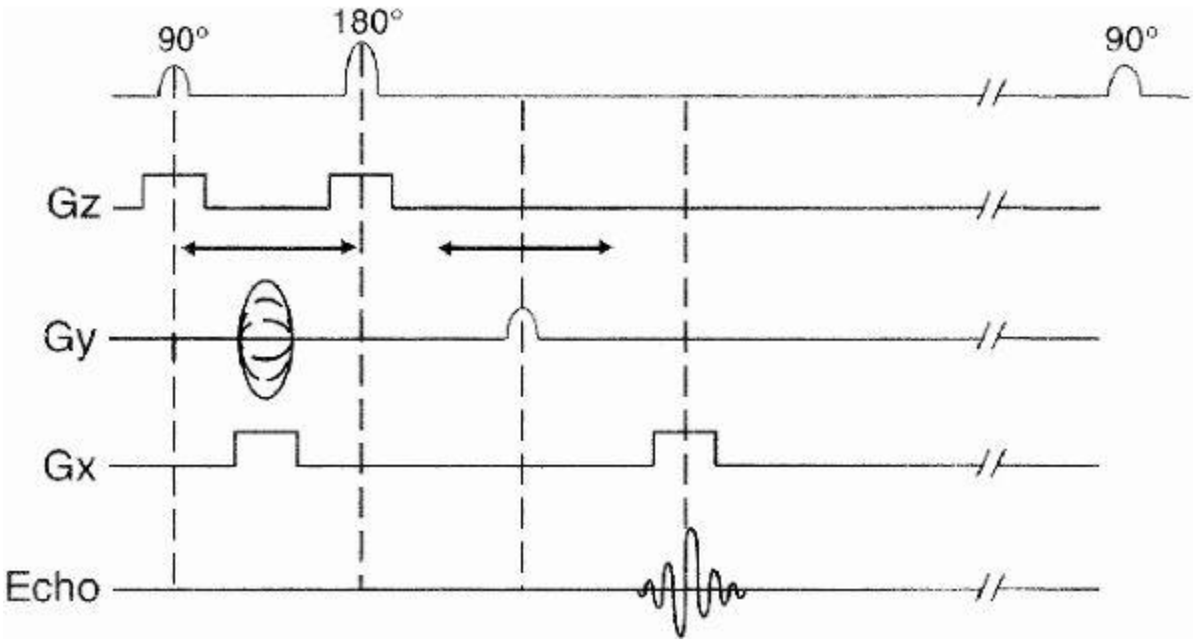
- Can we apply frequency encoding in 2 directions simultaneously?
  - Answer is NO
- 2D Fourier transform

$$\boxed{F(k_x, k_y) = \iint f(x, y) \cdot e^{-j2\pi(k_x \cdot x + k_y \cdot y)} dxdy}$$

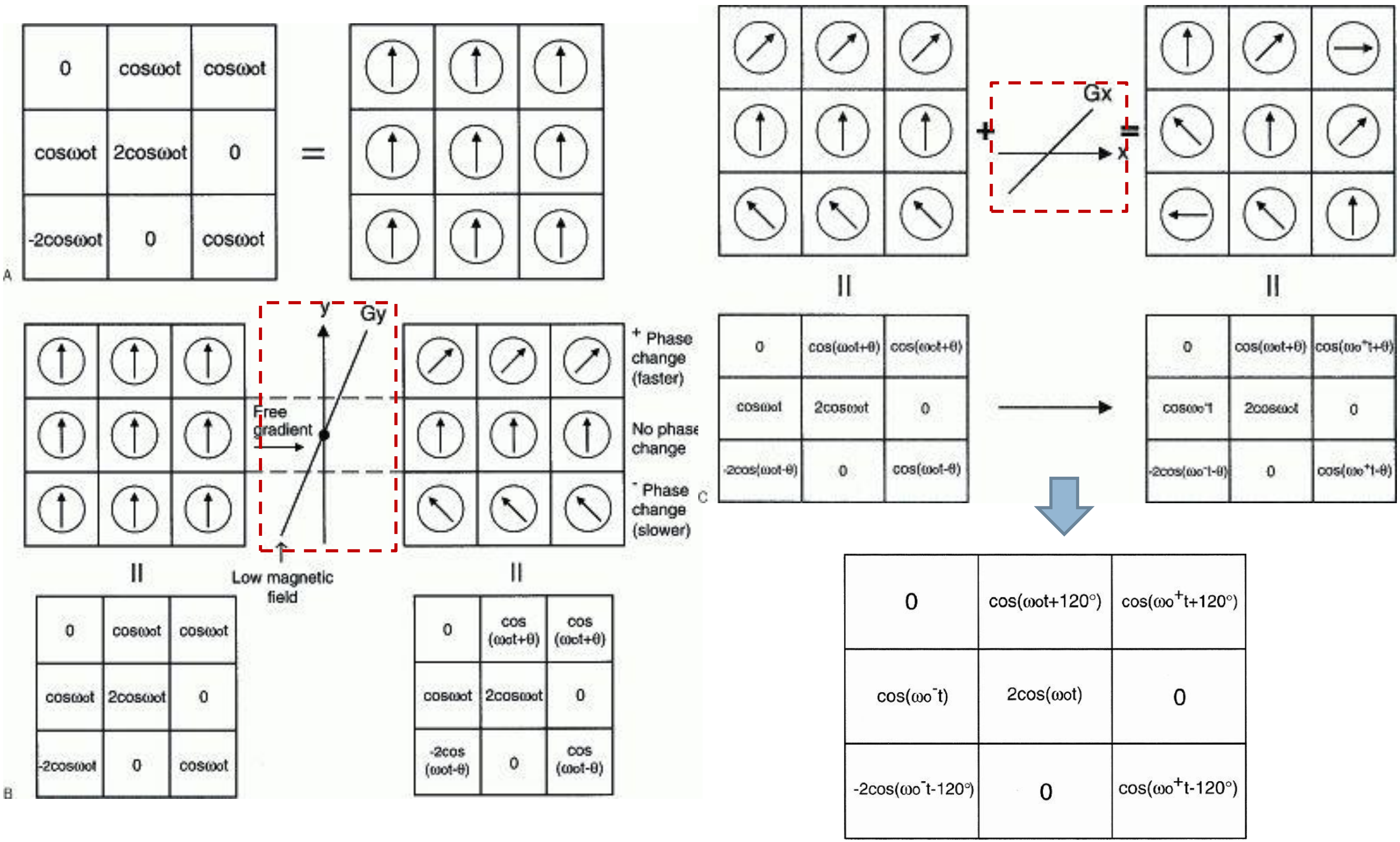
$$F(k_x, k_y) = \int \underbrace{e^{-j2\pi k_y y}}_{\text{Phase Encoding}} \underbrace{\left\{ \int f(x, y) \cdot e^{-j2\pi k_x x} dx \right\}}_{\text{Frequency Encoding}} dy$$

Phase Encoding

- G_y is usually applied between the 90° and the 180° RF pulses or between the 180° pulse and the echo.

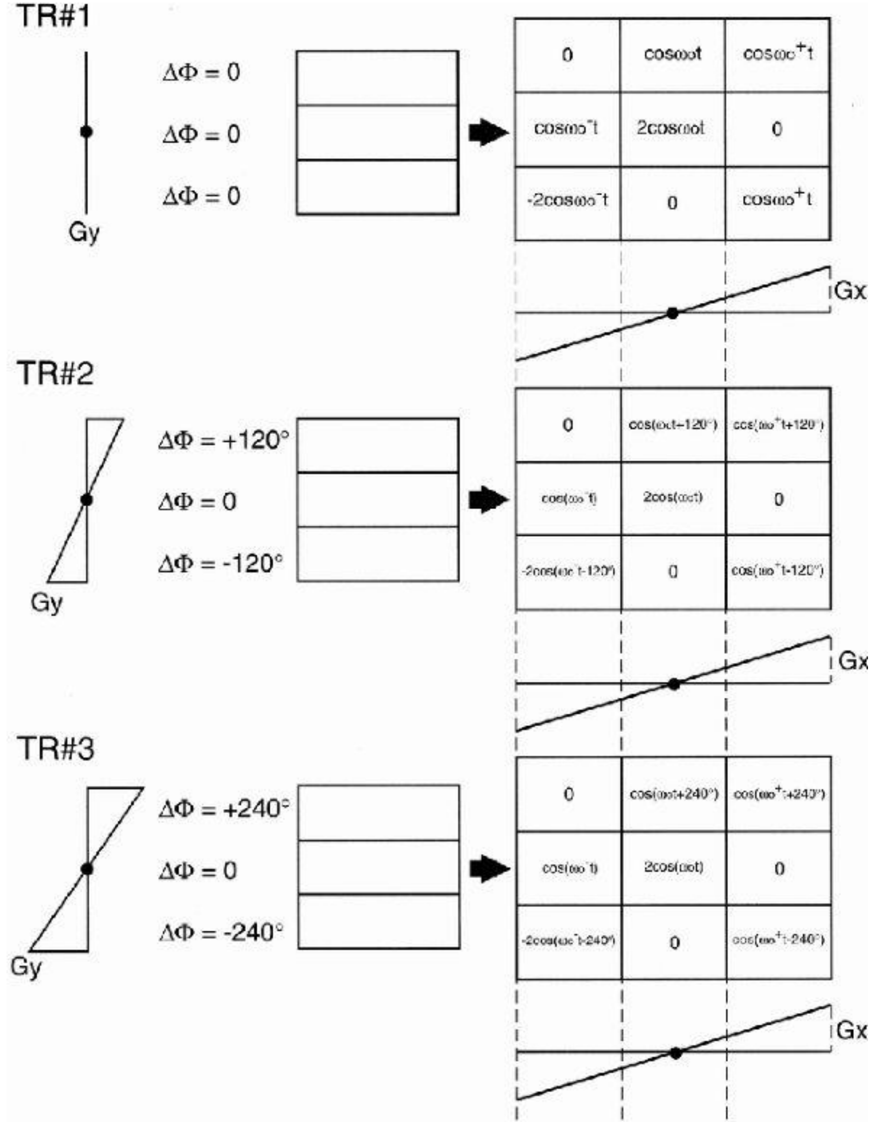


Phase Encoding



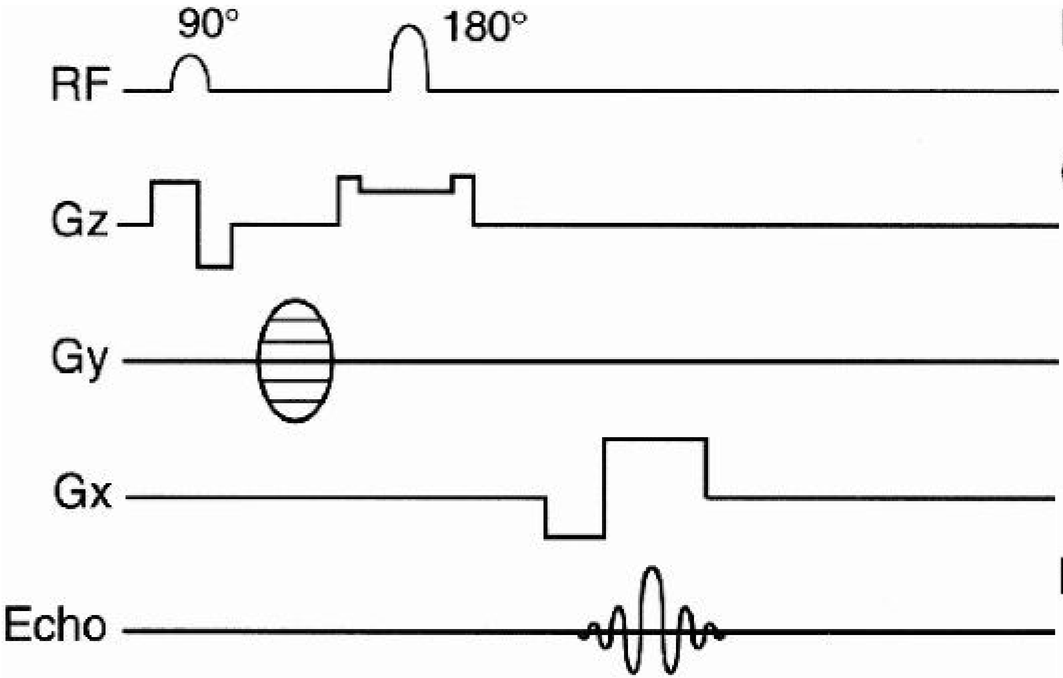
Phase Encoding

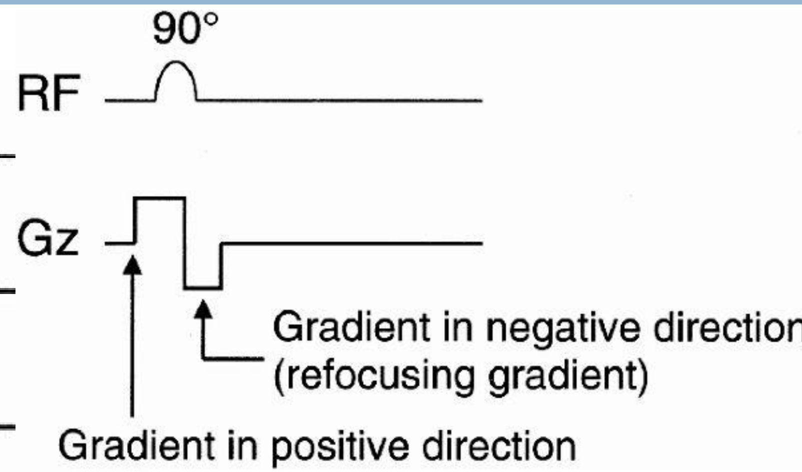
- Each phase encoding requires 1 RF pulse
  - Acquisition time = #phase encoding steps x TR

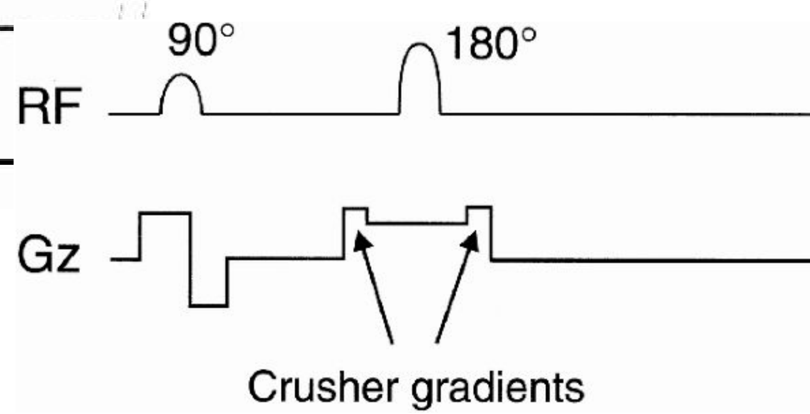


The protons in each pixel have a distinct frequency and a distinct phase, which are unique and encode for the x and y coordinates for that pixel.

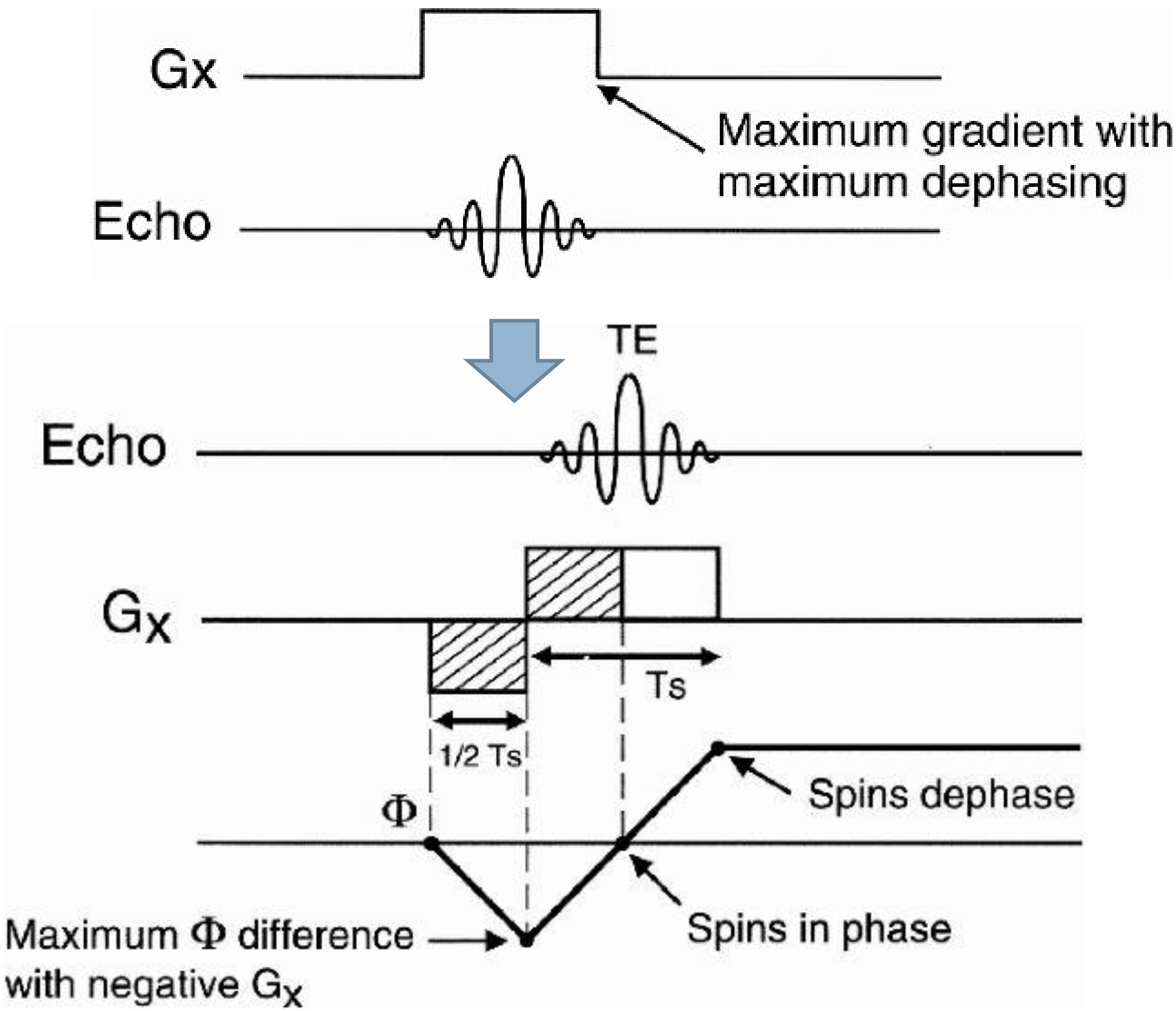
Pulse Sequence Diagram







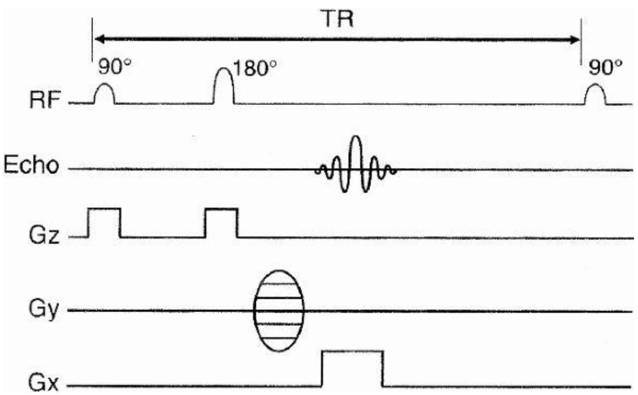
Pulse Sequence Diagram

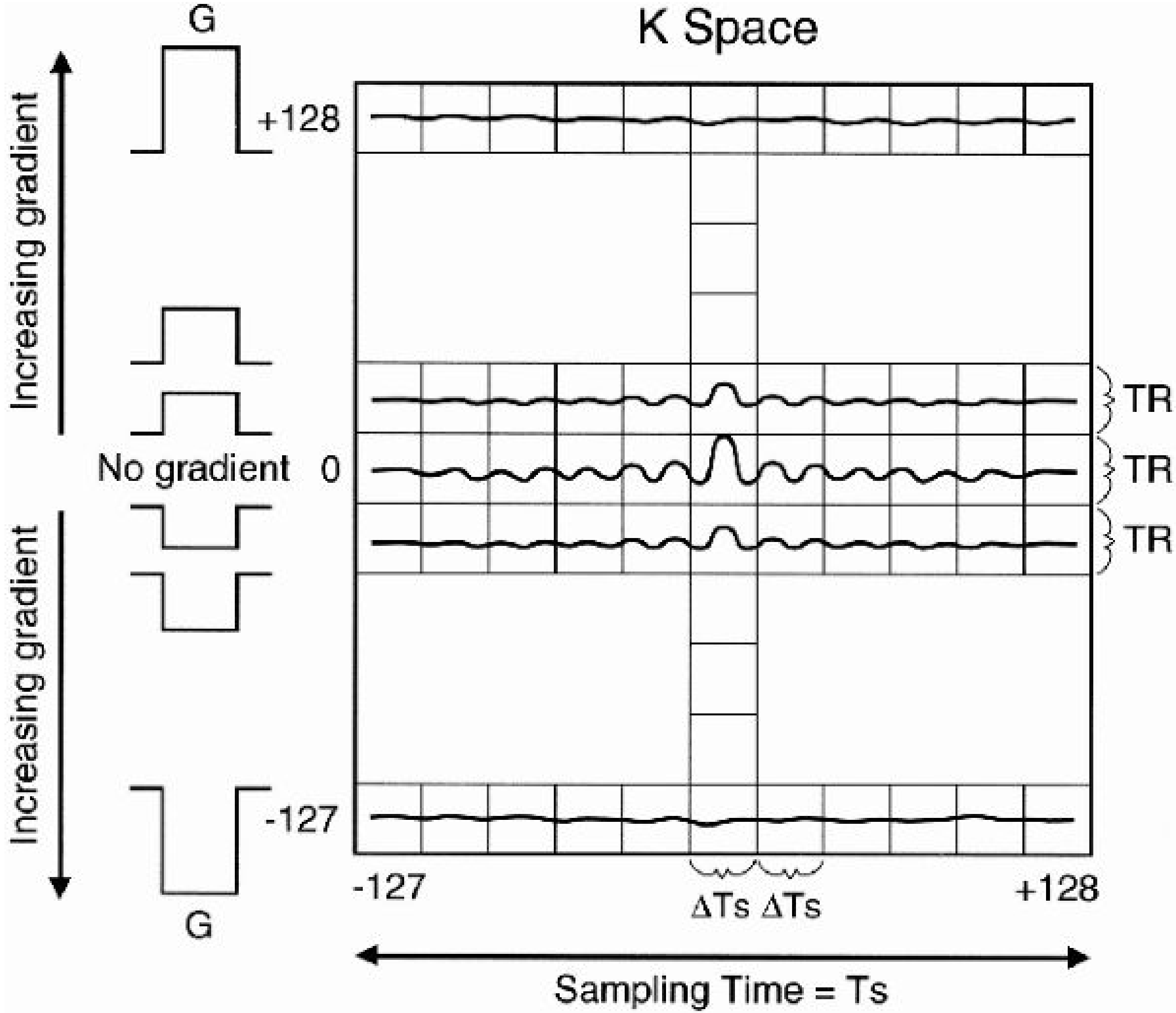


K-Space and Image Space

- K-space= Fourier domain

Sample Spin-Echo Pulse Sequence

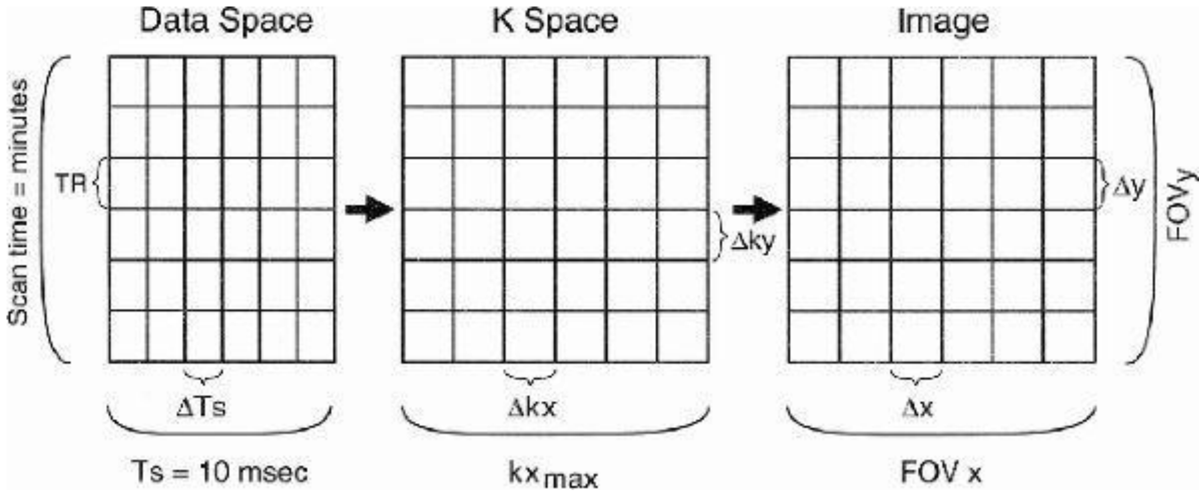


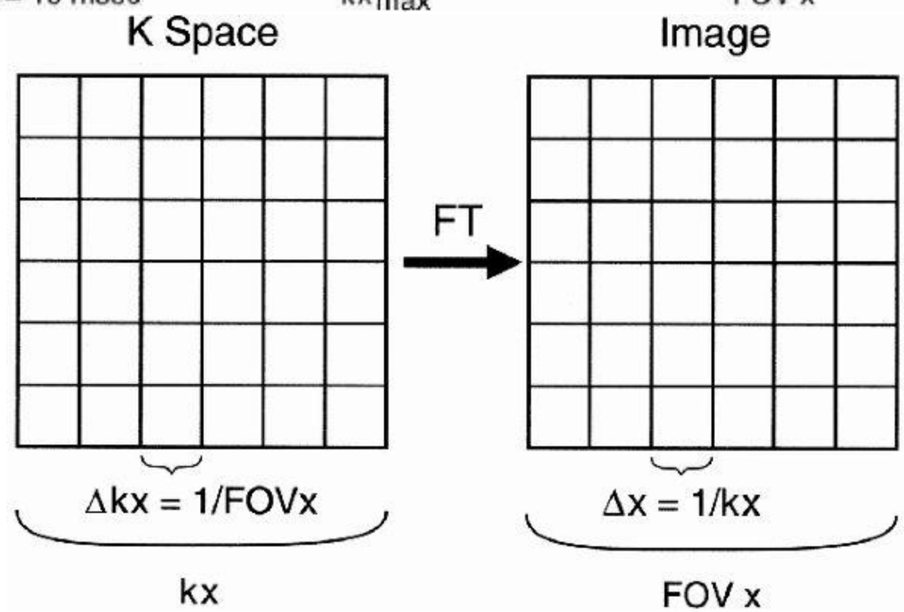


K-Space and Image Space

- Spatial frequencies k_x and k_y are expressed as:
  - $k_x = \gamma \int_0^t G_x(\tau)d\tau$
  - $k_y = \gamma \int_0^t G_y(\tau)d\tau$

with units in cycles/cm.





MRI Scan Parameters

Primary:

TR
TE
TI
FA (flip angle)

contribute to *image contrast*

Δz = slice thickness
Interslice gap

contribute to *coverage*

FOV_x
FOV_y

N_x : # of frequency-encoding steps
N_y : # of phase-encoding steps

Contribute to *resolution:*
Δx : spacing in x direction
Δy : spacing in y direction

NEX
Bandwidth

Contribute to *S/N ratio*

Secondary:

- SNR
- Scan time
- Coverage
- Resolution
- Image contrast

Parameter Optimization

- **SNR** defines as ratio of signal magnitude to noise standard deviation
  - Voxel volume = $\Delta x \cdot \Delta y \cdot \Delta z$
  - Number of excitations (NEX)
  - Number of phase-encoding steps (N_y and N_z)
  - Bandwidth (BW)

$$\boxed{3D\ SNR \propto \Delta x \cdot \Delta y \cdot \Delta z \sqrt{(N_y)(N_z)(NEX)/BW}}$$

- SNR can be increased by
  - Increasing TR
  - Decreasing TE
  - Using a lower BW
  - Using volume (i.e., 3D) imaging
  - Increasing NEX
  - Increasing N_y
  - Increasing the voxel size

Parameter Optimization

- **Spatial resolution** (or pixel size) is the minimum distance that we can distinguish between two points on an image.
- It is determined by Pixel size = FOV/# of pixels
  - For example, pixel size in y = $\Delta y = FOV_y/N_y$
  - Nx, Ny, Nz are called **Matrix Size**
- If we want higher resolution in a given time, we have to sacrifice SNR:

$$\text{SNR (3D)} = (\text{FOV}_x/\text{N}_x)(\text{FOV}_y)(\text{FOV}_z)\sqrt{\frac{\text{NEX}}{(\text{N}_y)(\text{N}_z)(\text{BW})}}$$

Parameter Optimization

- **Acquisition Time** or **Scan Time**, as we have seen previously, is given by

$$\text{Scan time} = TR \cdot N_y \cdot N_z \cdot NEX$$

where Ny, Nz are the number of phase-encoding steps (in the y and z directions)

- If we have a multi-slice sequence (i.e., no phase encoding in z direction), then we may be able to squeeze in each TR multiple slice acquisition
  - Maximum of TR/TE slices

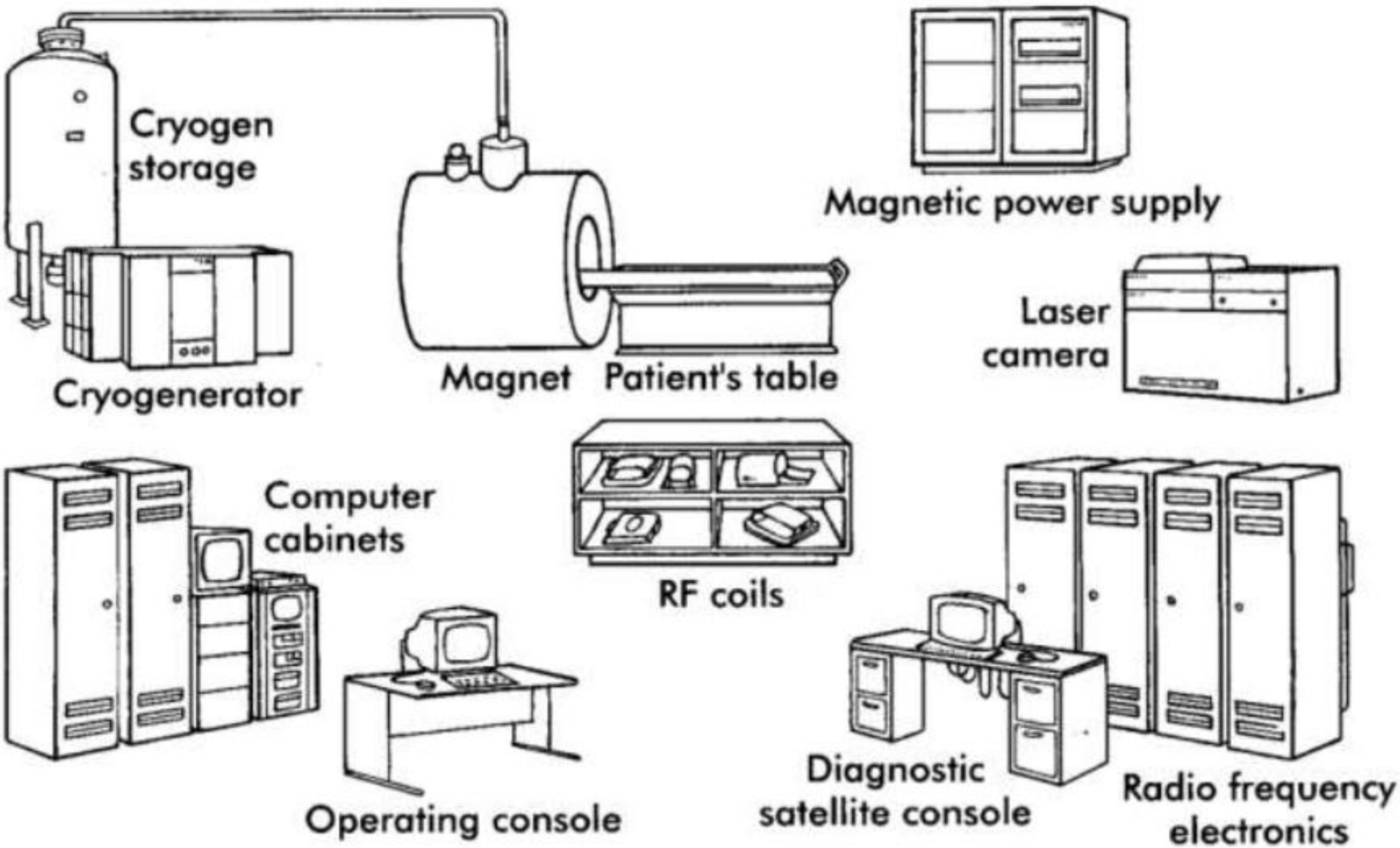
Parameter Optimization: Examples

- If we keep FOV constant and increase N_y, we will decrease SNR. $\uparrow N_y$, FOV constant $\rightarrow \downarrow$ SNR
- If we increase N_y and increase FOV, thus keeping pixel size constant, then we will increase the SNR.
  - $\uparrow$ FOV, pixels fixed $\rightarrow \uparrow$ SNR, $\uparrow$ acquisition time
- If we increase the number of pixels with the FOV constant:
  - Increase resolution.
  - Decrease SNR Therefore, as we decrease the pixel size, we increase the resolution and decrease the SNR.
  - Increase scan time (number of pixels increases in phase-encode direction).

Parameter Optimization: Examples

- if we decrease the FOV and keep number of pixels constant:
  - Increase the resolution.
  - Decrease SNR.
  - Same acquisition time
- In the x direction, there are two ways of increasing resolution (for a given FOV):
  - Increase N_x by reducing the sampling time ΔTs (i.e., by increasing the BW) and keeping the total sampling time Ts fixed (recall that $Ts = N_x \cdot \Delta Ts$). The advantage here is no increase in TE; the trade-off is a reduction in SNR (due to increased BW).
  - Increase N_x by lengthening Ts and keeping ΔTs (and thus BW) fixed. Here, the SNR does not change, but the trade-off is an increased TE (due to a longer Ts) and less T1 weighting (this is only a concern in short echo delay time imaging).

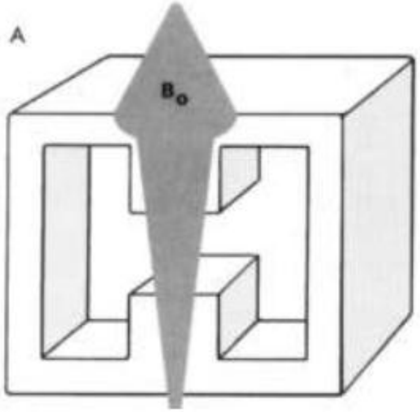
Block Diagram of MRI System

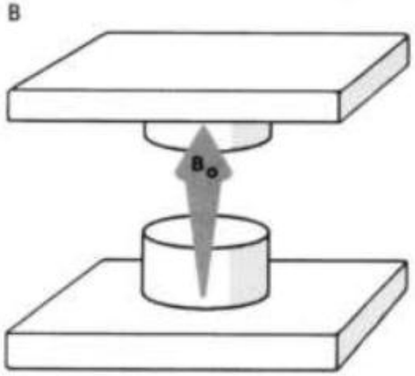


Primary Magnetic Field (B0)

- Permanent magnet
- Resistive magnet
- Superconductive magnet

Permanent Magnet





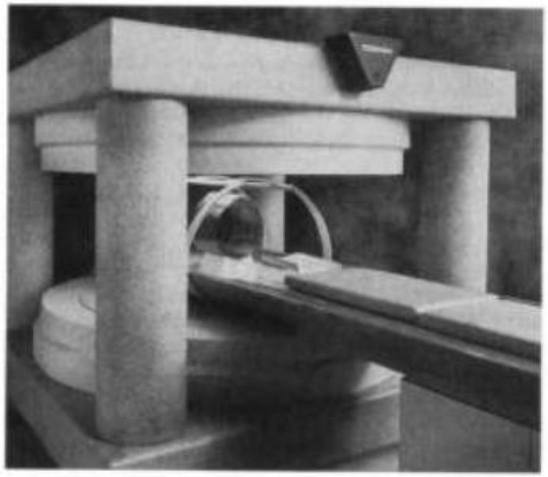


Table 11-1 Characteristics of a permanent magnet magnetic resonance imager

Feature	Value
Magnetic field (B_o)	Up to 0.3 T
Magnetic field homogeneity	50-100 ppm
Weight	90,000 kg
Cooling	None
Power consumption	20 kW
Distance to 0.5 mT fringe field	< 1 m

Resistive Magnet

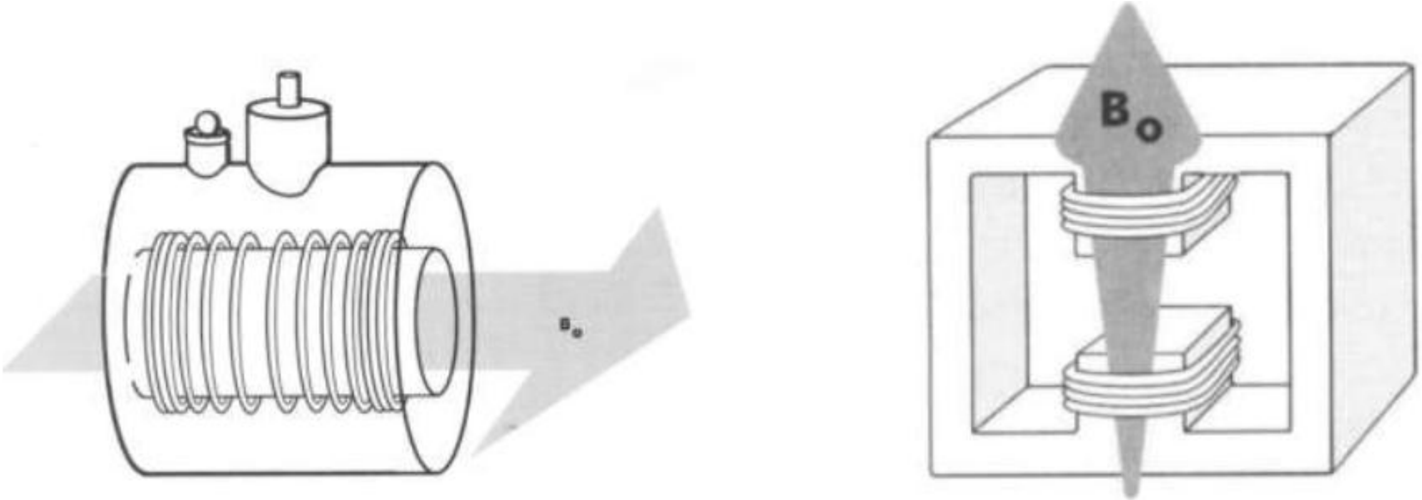
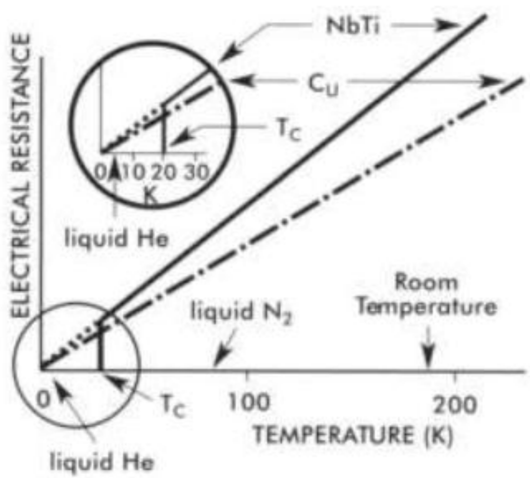


Table 11-2 Characteristics of a resistive electromagnet MR imager

Feature	Value
Magnetic field (B_o)	Up to 0.3 T
Magnetic field homogeneity	10-50 ppm
Weight	4000 kg
Cooling	Water, heat exchanger
Power consumption	80 kW
Distance to 0.5 mT fringe field	2 m

Superconductive Magnet



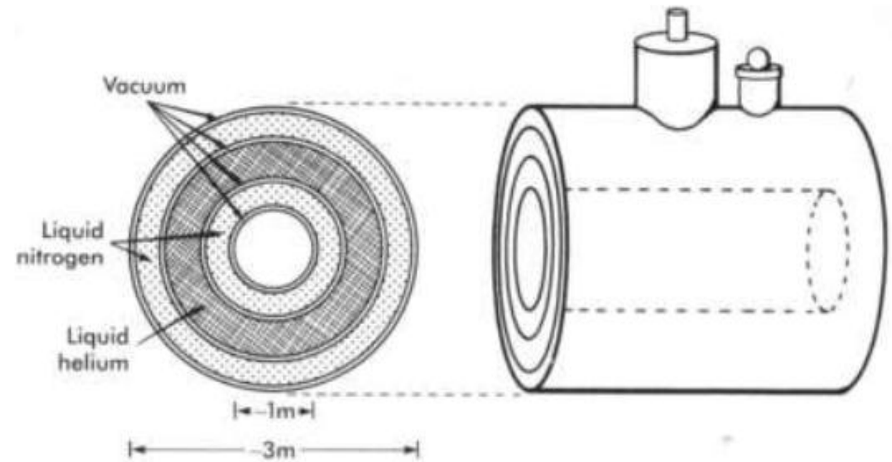
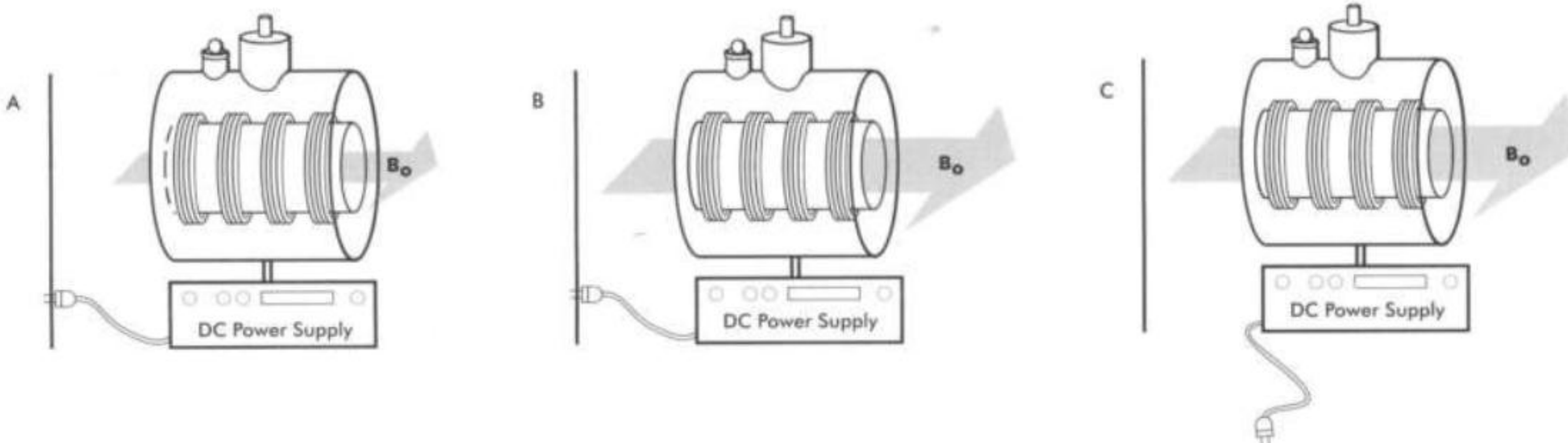


Table 11-3 **Characteristics of a superconducting electromagnet magnetic resonance imager**

Feature	Value
Magnetic field (B_o)	0.3 T to 4 T
Magnetic field homogeneity	1-10 ppm
Weight	10,000 kg
Cooling	Cryogenic
Power consumption	20 kW
Distance to 0.5 mT fringe field	10 m

Superconductive Magnet

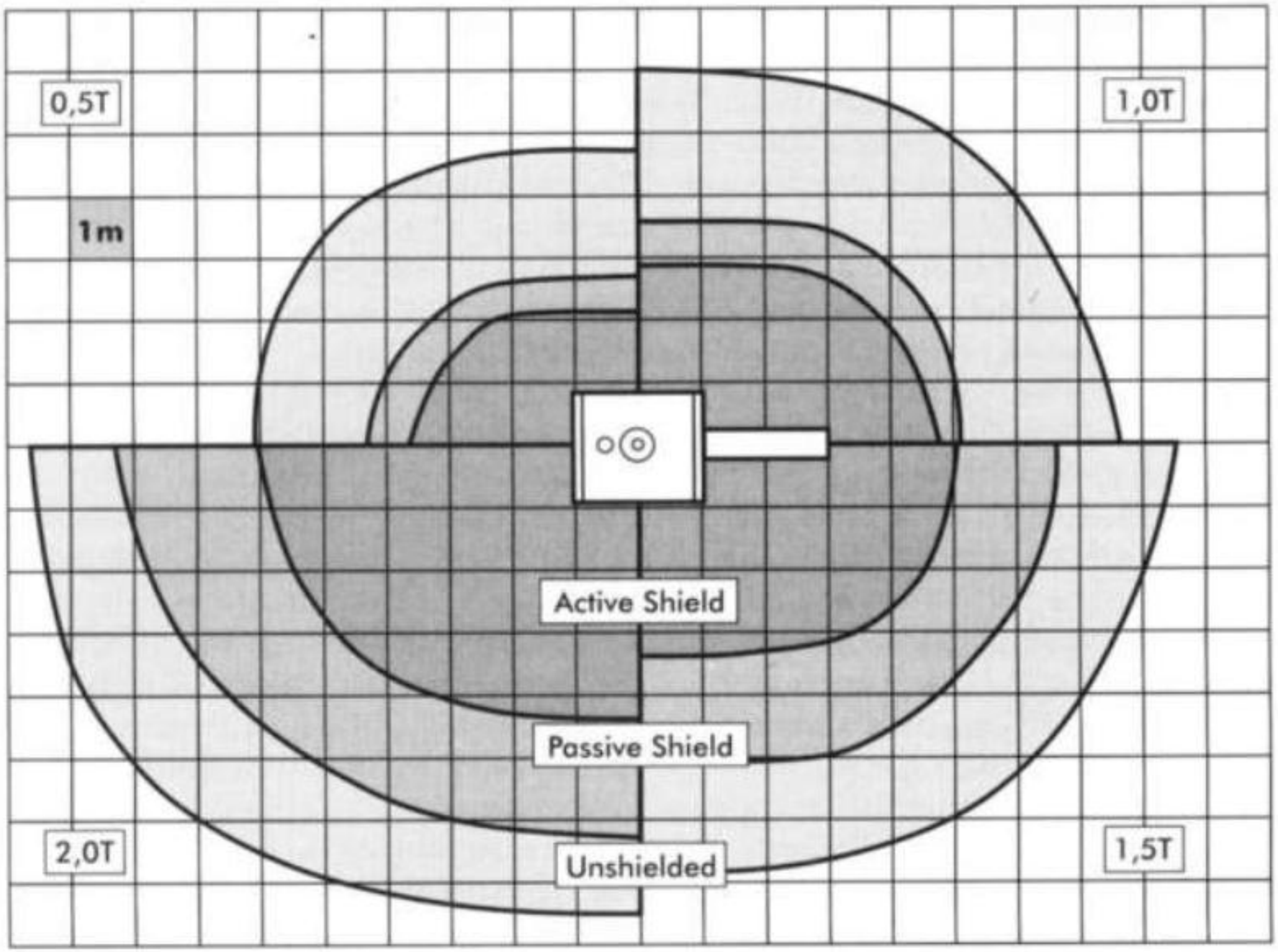
- Magnetic field ramp-up



- Ramp-down must be very slow, otherwise catastrophic quenching will occur
  - Heating up increases resistance, which in turn increases heating, causing positive feedback loop that can result in rapid vaporization of helium

Magnet Shielding

- None
- Passive
- Active



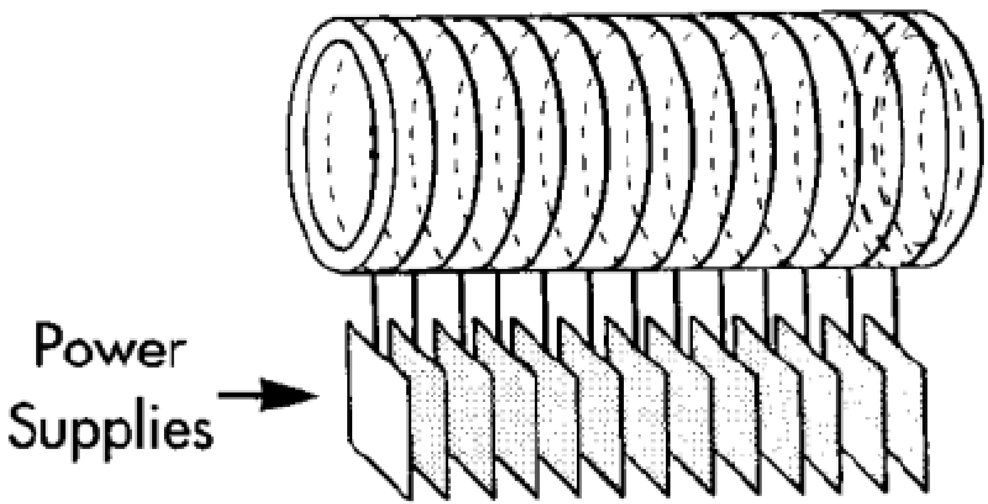
Distance to Safe 5G (0.5 mT) Line for different fields

Secondary Magnets: Coils

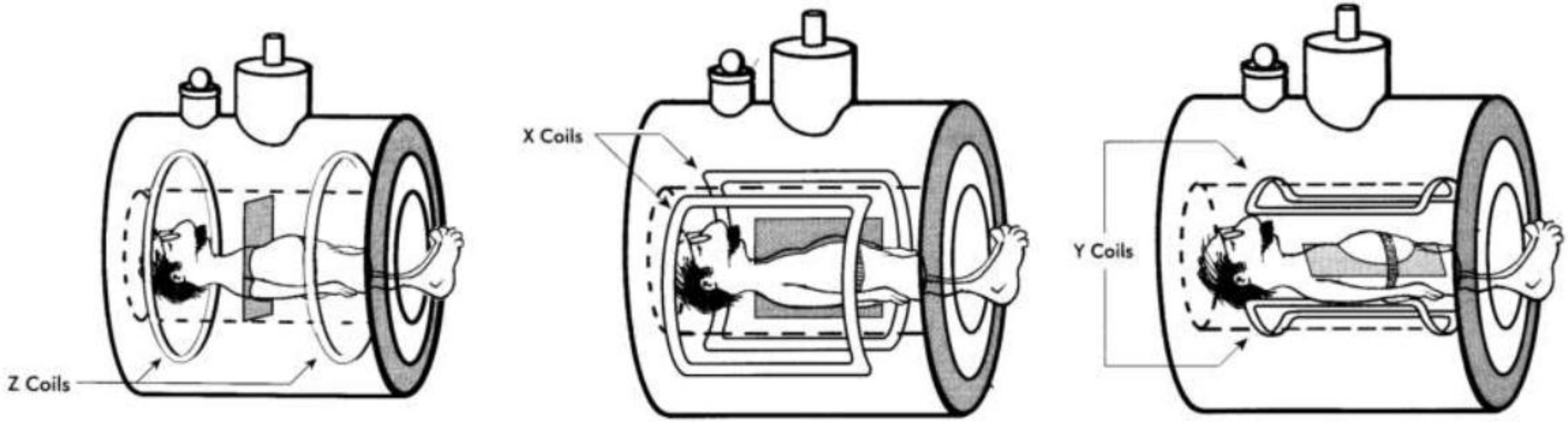
- Shim coils
  - Improves B0 field uniformity to within a few **ppm scale**
- Gradient coils
  - Apply gradients in x, y, and z directions for slice selection, frequency and phase encoding.
- RF coils
  - Send RF pulses and receive signal from patient

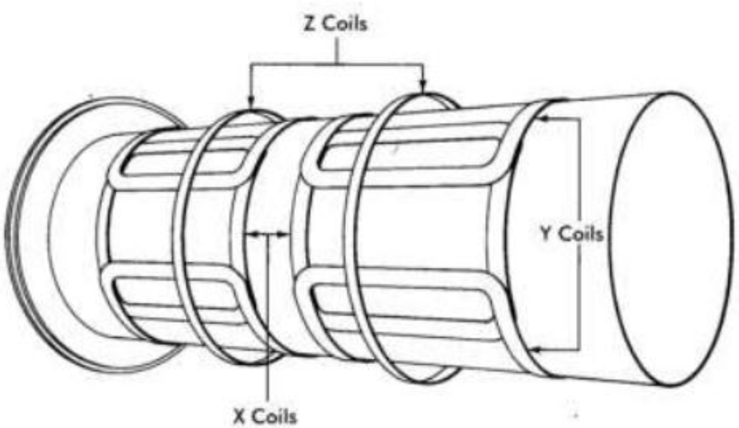
Shim Coils

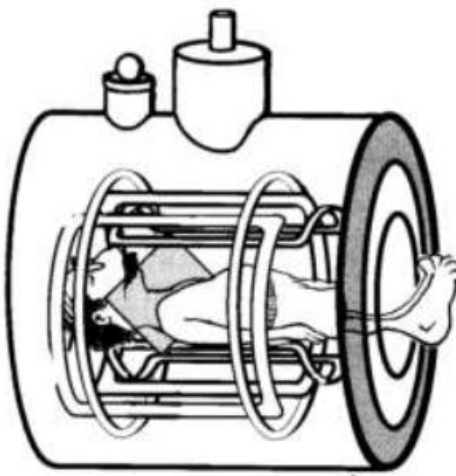
- Make small adjustments to make B0 uniform throughout the volume
  - Inhomogeneity measured in ppm units
  - Example: for 1.0T magnet, a homogeneity of $\pm$1ppm means that the field has a variation of up to $\pm 1\ \mu T$



Gradient Coils

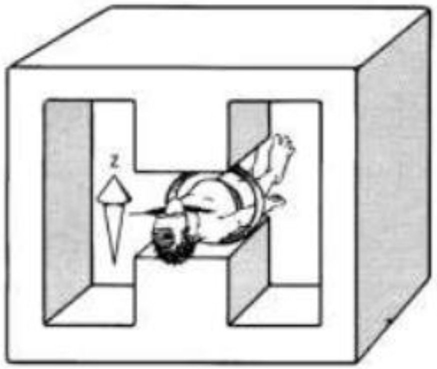


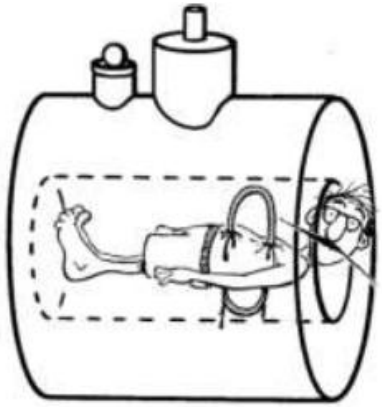


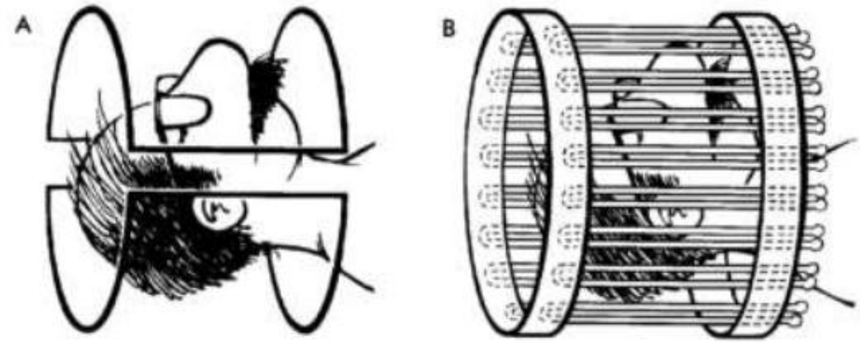


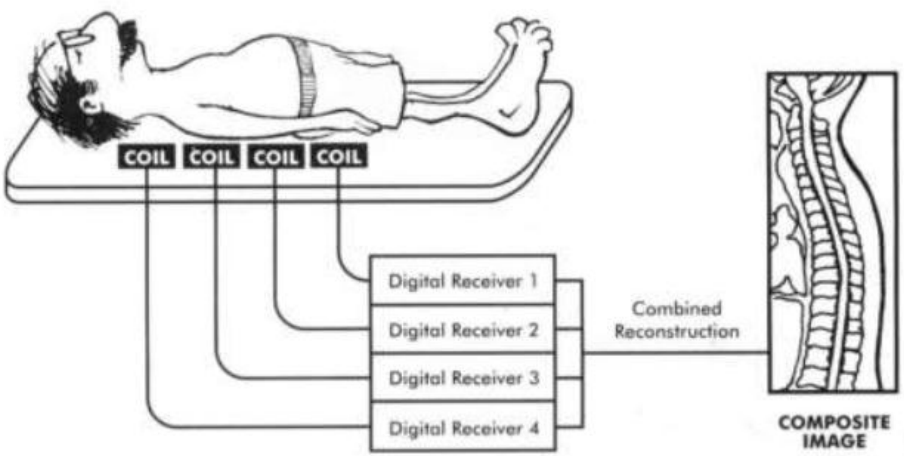
- Units: mT/m
- Slew rate
- Limited for safety

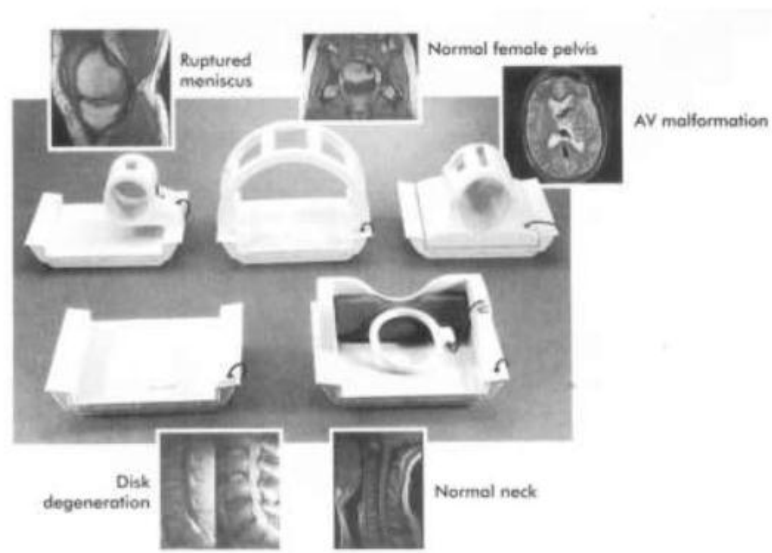
RF Coils






A
B

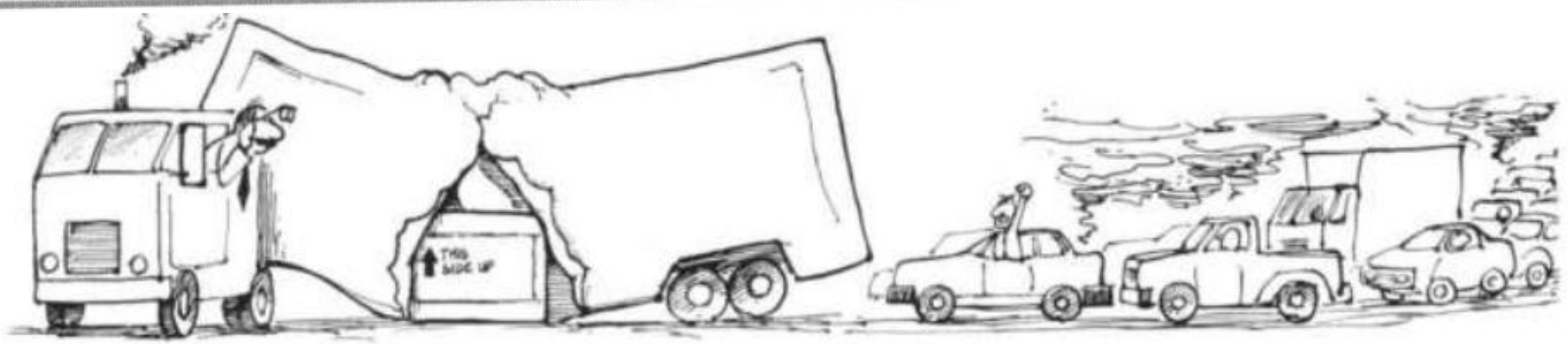

COIL
COIL
COIL
COIL
Digital Receiver 1
Digital Receiver 2
Digital Receiver 3
Digital Receiver 4
Combined Reconstruction
COMPOSITE IMAGE


Ruptured meniscus
Normal female pelvis
AV malformation
Disk degeneration
Normal neck

Choosing a Magnet Type

Table 13-1 Characteristics of magnetic resonance imagers

Characteristics	Permanent magnet	Resistive magnet	Superconducting magnet
Field strength (T)	0.1-0.3	0.15-0.4	0.5-4.0
Cost ($\$ \times 10^6$)	0.5-1.0	0.8-1.2	1.0-2.5
Approximate size (m)	1.5×2.0	2.1×2.3	2.3×3.0
Weight (kg $\times$ 1000)	4.5-30	5.5-9.0	4.5-8.1
Power requirements (kW)	20	80	25
Distance to 0.5 mT fringe field (m)	<1	0.5-2	3-10



Choosing a Magnet Type

Table 13-2 Advantages and disadvantages of magnetic resonance imagers

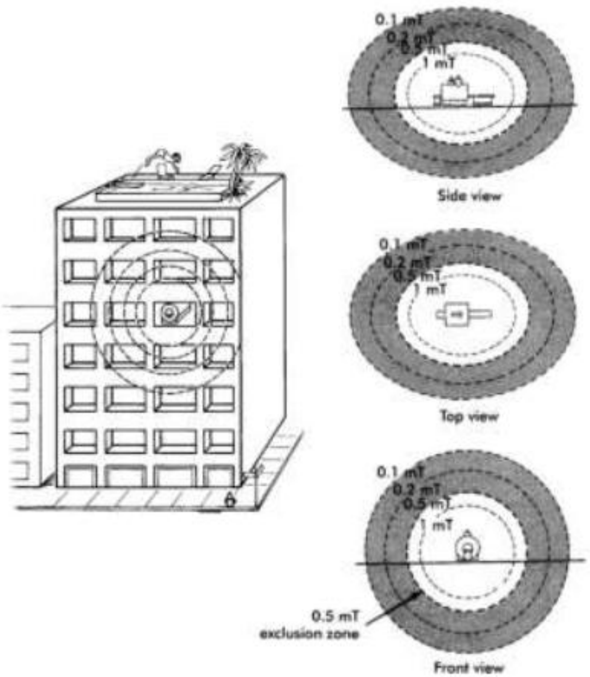
Advantages	Disadvantages
Permanent	
Low capital cost	Limited field strength
Low operating cost	Fixed field strength
Negligible fringe field	Very heavy
Resistive Iron Core	
Low capital cost	High power consumption
Easy coil maintenance	Water cooling necessary
Negligible fringe field	Potential field instability
Resistive air core	
Low capital cost	High power consumption
Lightweight	Water cooling necessary
Easy coil maintenance	Significant fringe field
Superconductive	
High field strength	High capital cost
High field homogeneity	High cryogen cost
Low power consumption	Intense fringe field

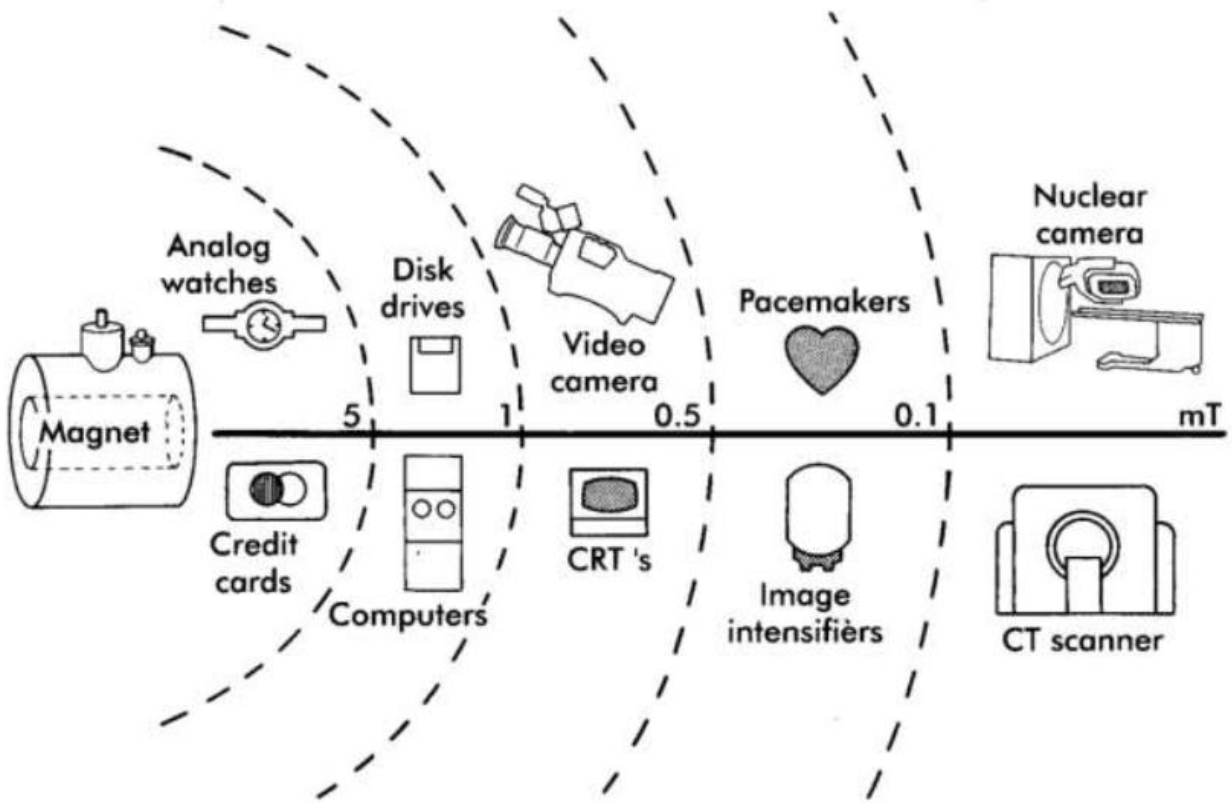
Site Selection for MRI

Table 13-3 Considerations for locating a magnetic resonance imager

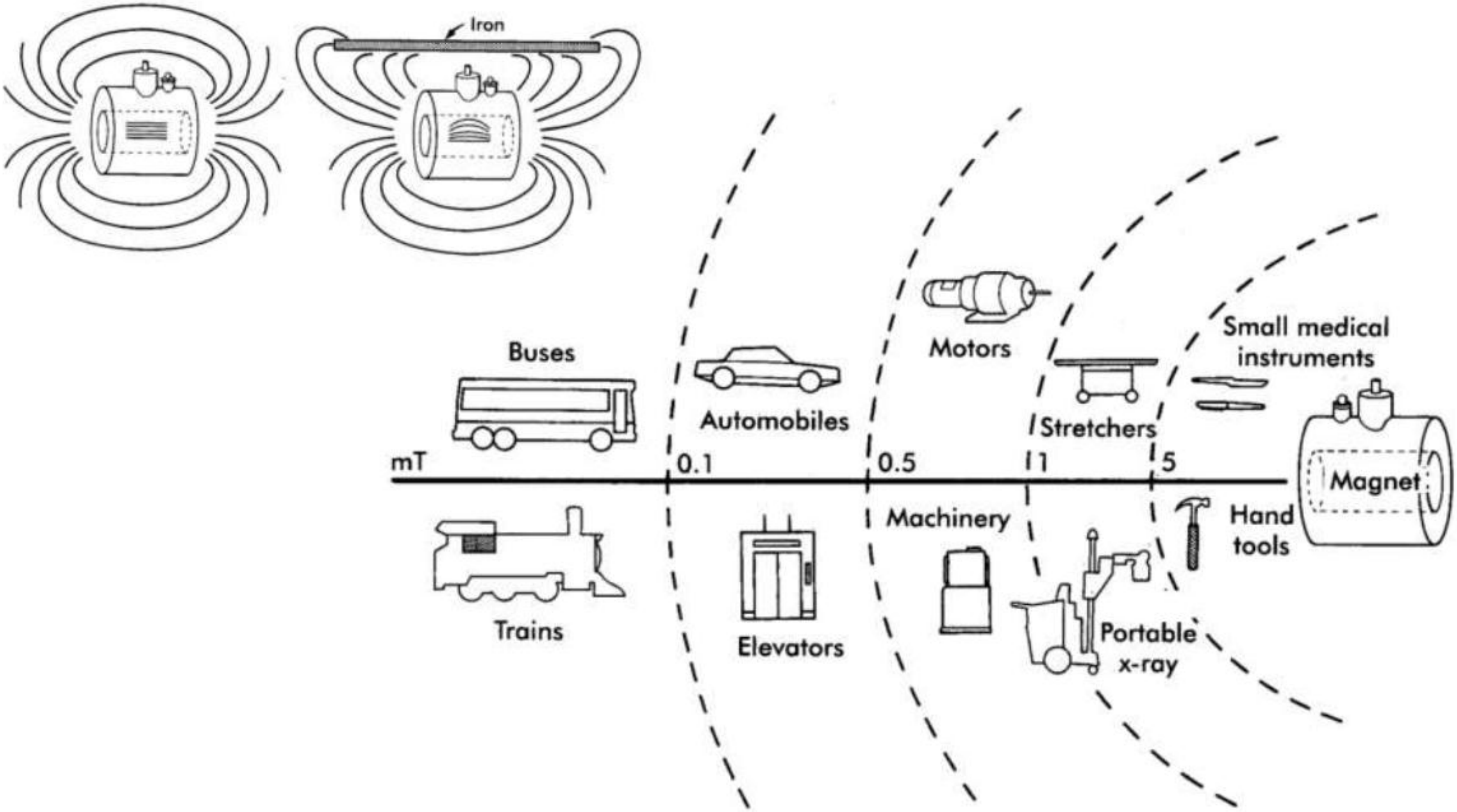
Advantages	Disadvantages
New construction	
Easier to plan for fringe magnetic field	Cost
Custom design	Possibly remote
Existing building	
Proximity to other services	Accommodation of fringe magnetic field, higher renovation cost
Use of existing facilities	
Temporary building	
Short time to operation	Possible compromised patient access
Easier to plan for fringe magnetic field	Unsightly addition
Mobile	
Cost effective for low workload	Scheduling
Learning period for all	Time required for setup

Effects of MRI on the Environment





Effects of the Environment on MRI



Covered Material

- Solve the problems at the end of Chapters 3, 4, 5, 6, 7, 8, 9, 10, 11, 13, 16 and 17.