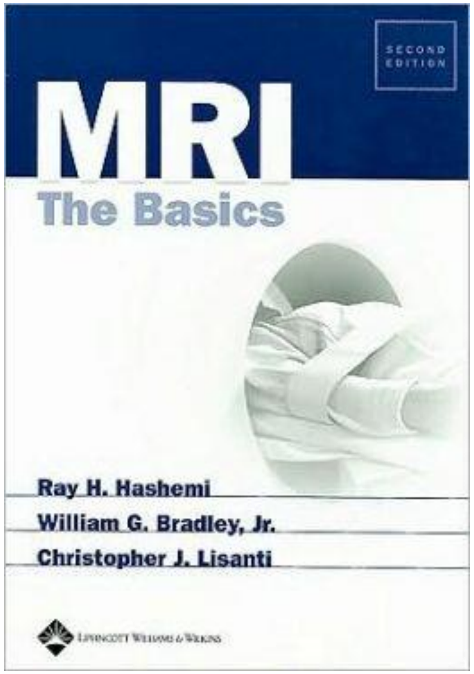
# MAGNETIC RESONANCE IMAGING

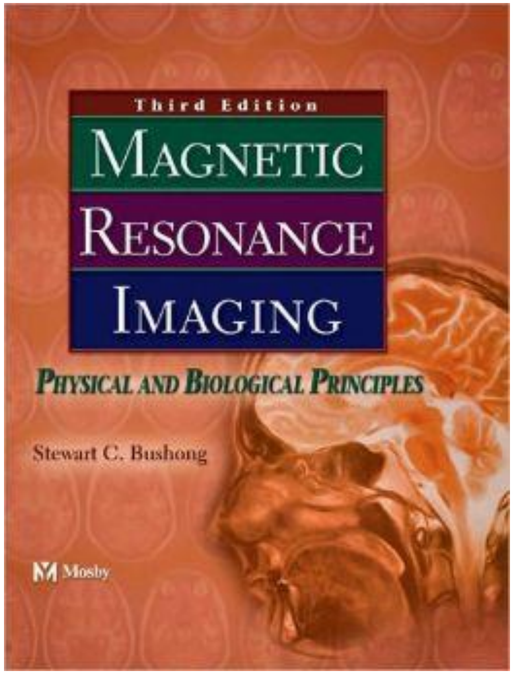
EE 472 – F2018

Prof. Yasser Mostafa Kadah – www.k-space.org

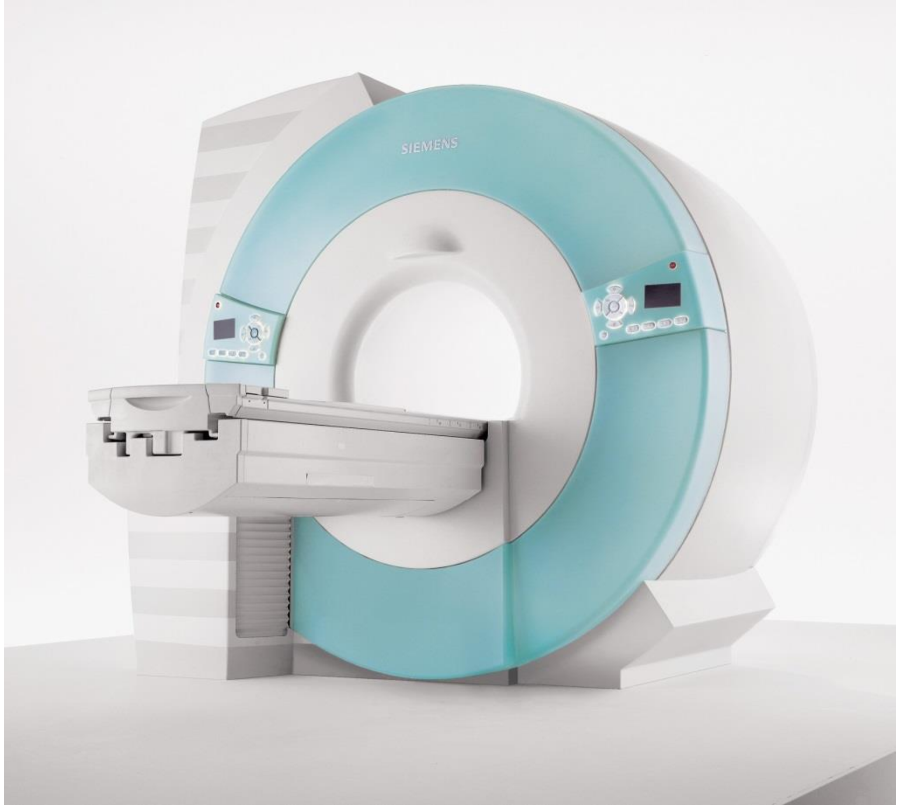
# Recommended Textbooks

- *MRI: The Basics, 2nd* Edition, by Ray H. Hashemi, William G. Bradley, and Christopher J. Lisanti, Lippincott Williams & Wilkins, 2003.
- *Magnetic Resonance Imaging: Physical and Biological Principles*, 3rd Edition, by Stewart C. Bushong, 2003.



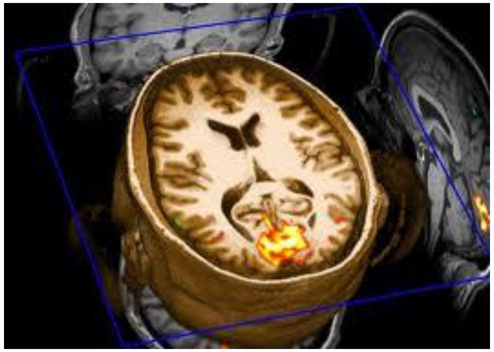
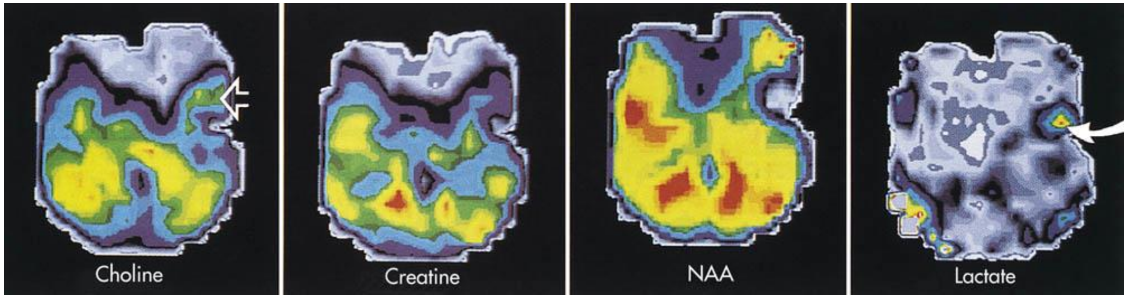
# Magnetic Resonance Imaging

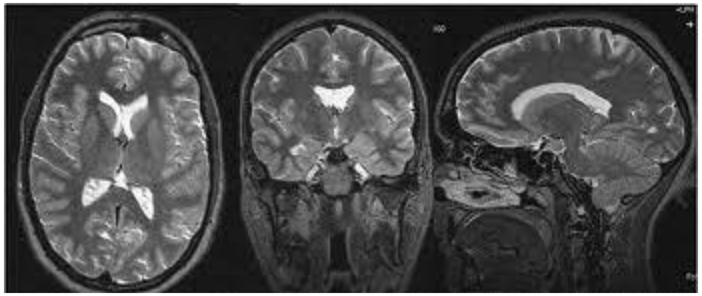



# Magnetic Resonance Imaging

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

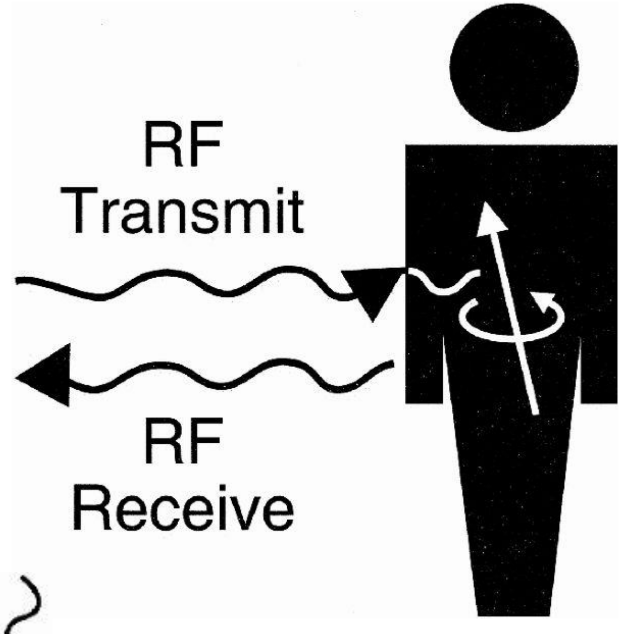


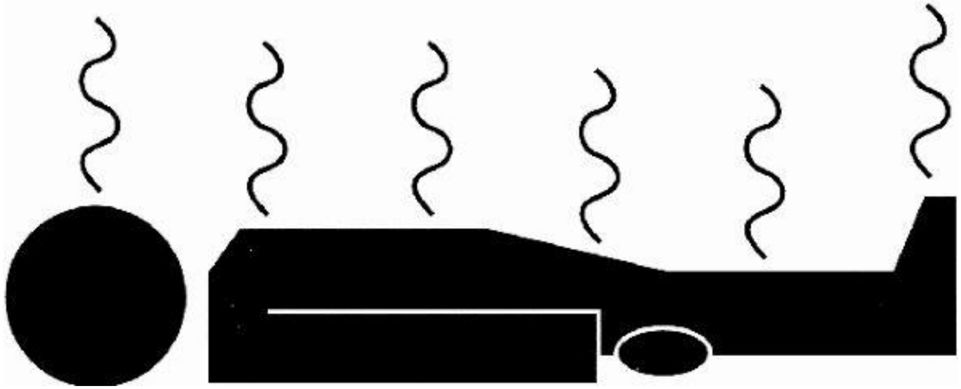






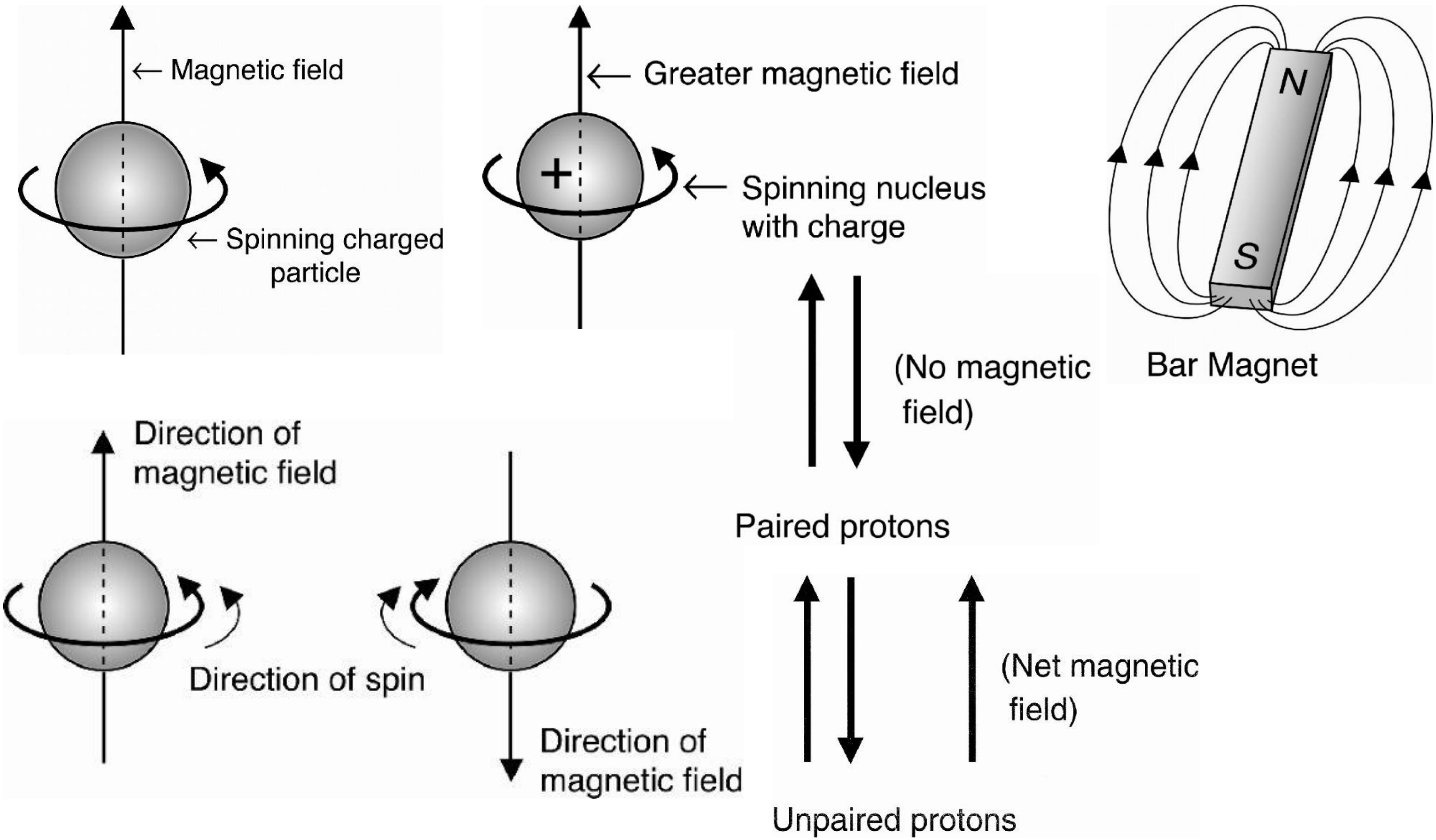
# Steps to Perform MR Imaging

- **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



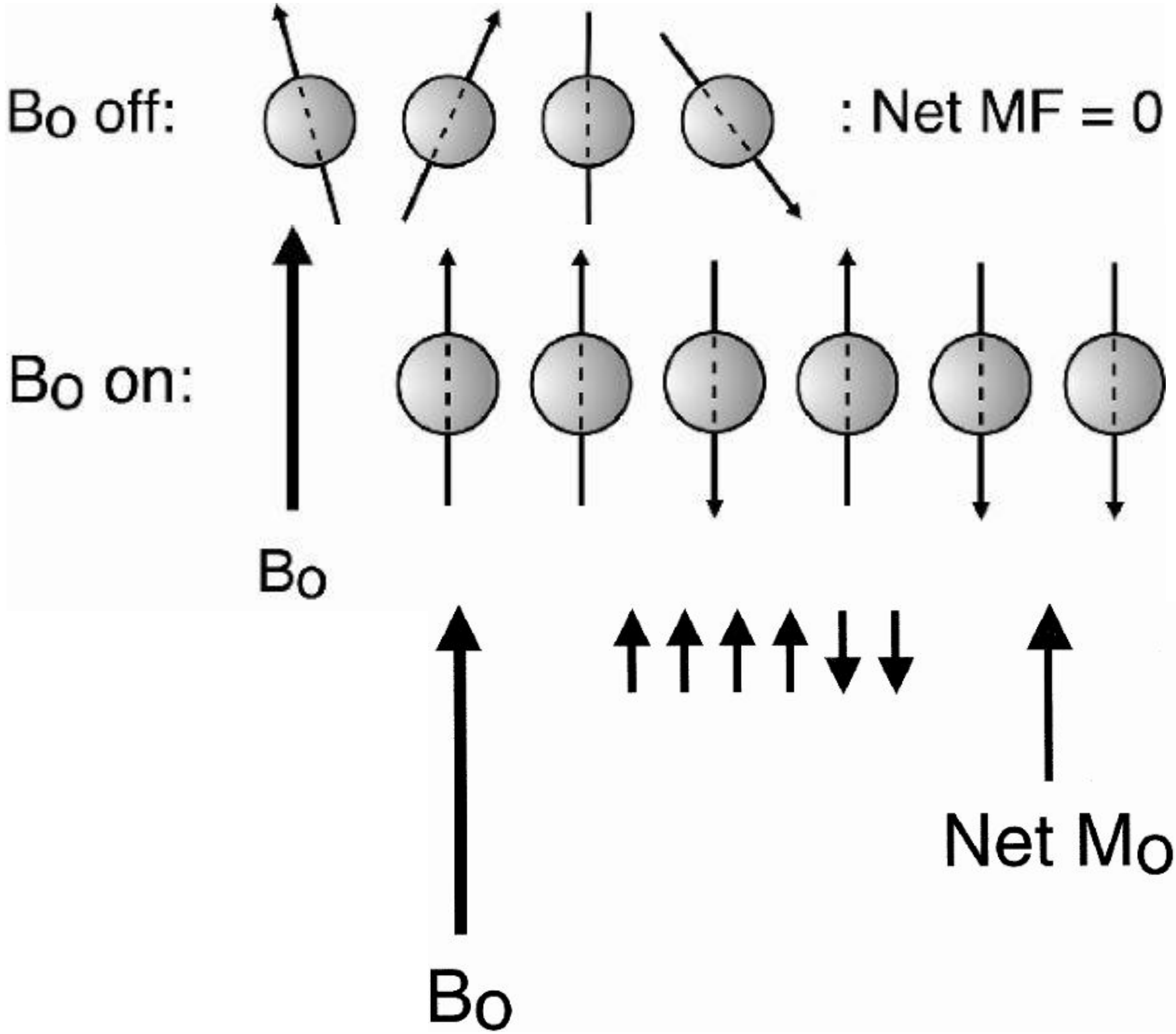



# Basic Physics



# 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 $e^{-(U/k_BT)} = e^{\gamma m\hbar B/k_BT}$

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

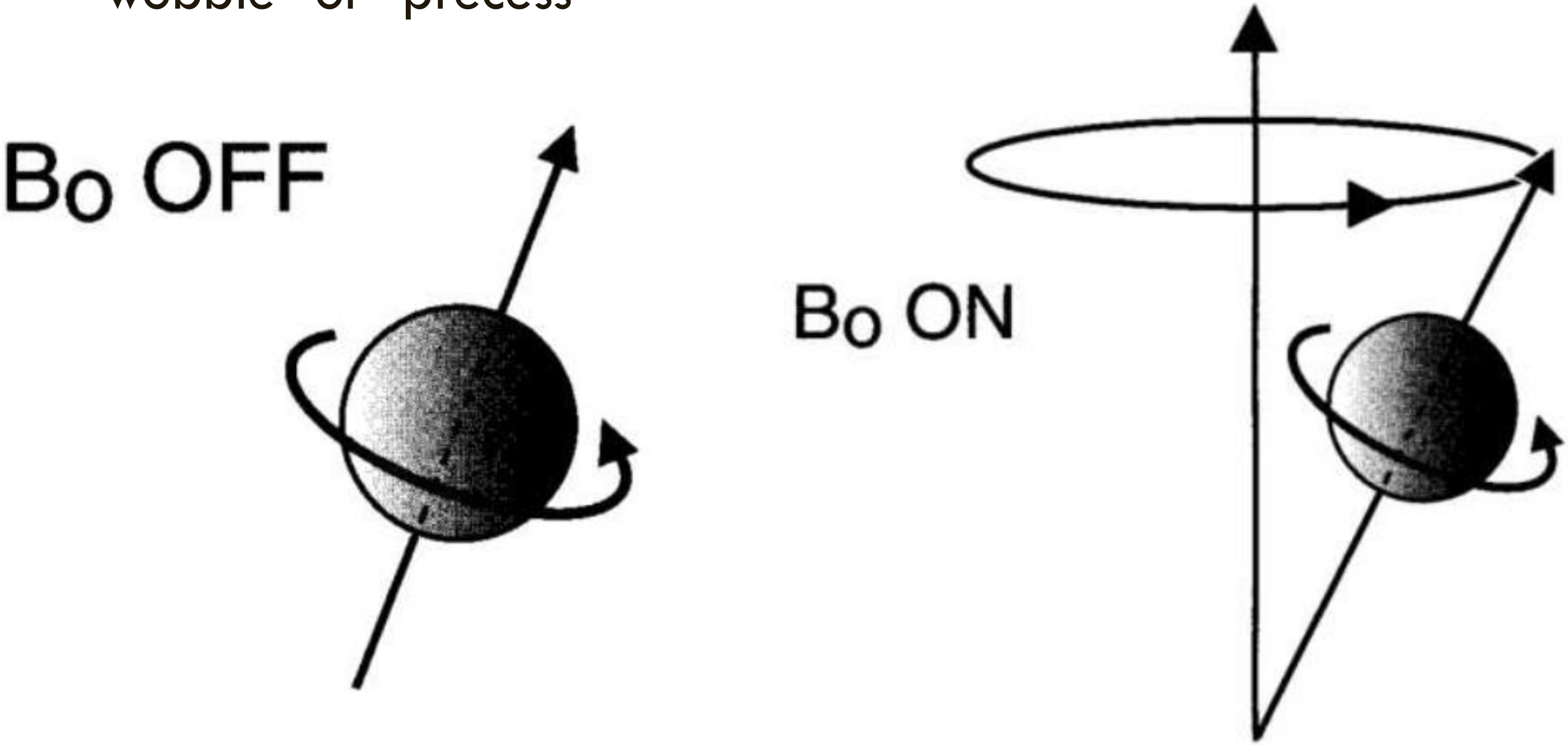
- At room temperature, $\gamma I\hbar B/k_BT \ll 1$

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

⇨ $M_z$ 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

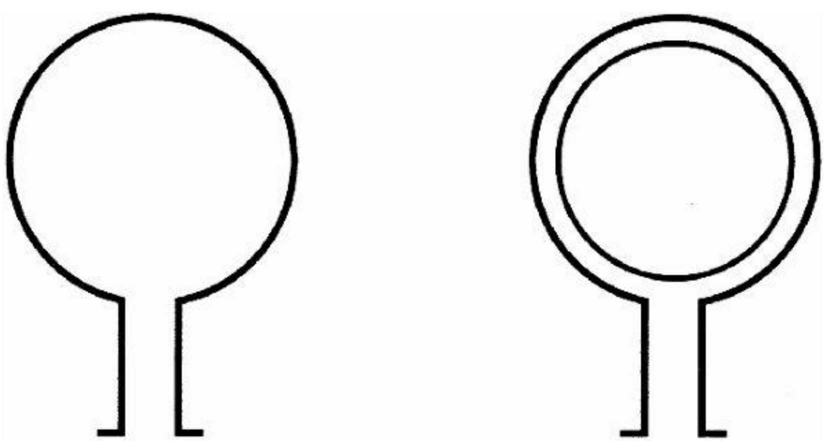
- The rate at which proton precesses around external magnetic field is given by:

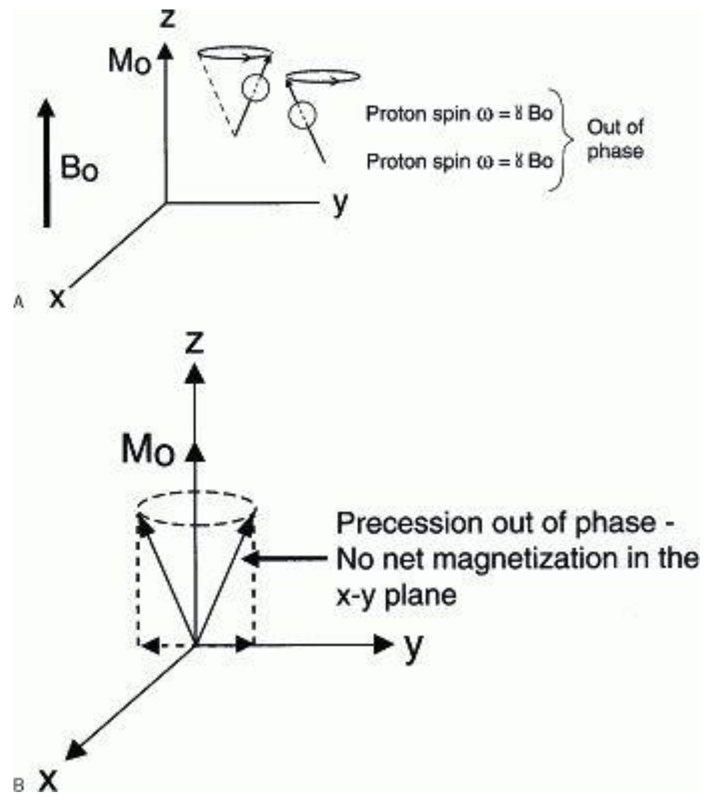
$$\omega = \gamma B_0$$

| 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

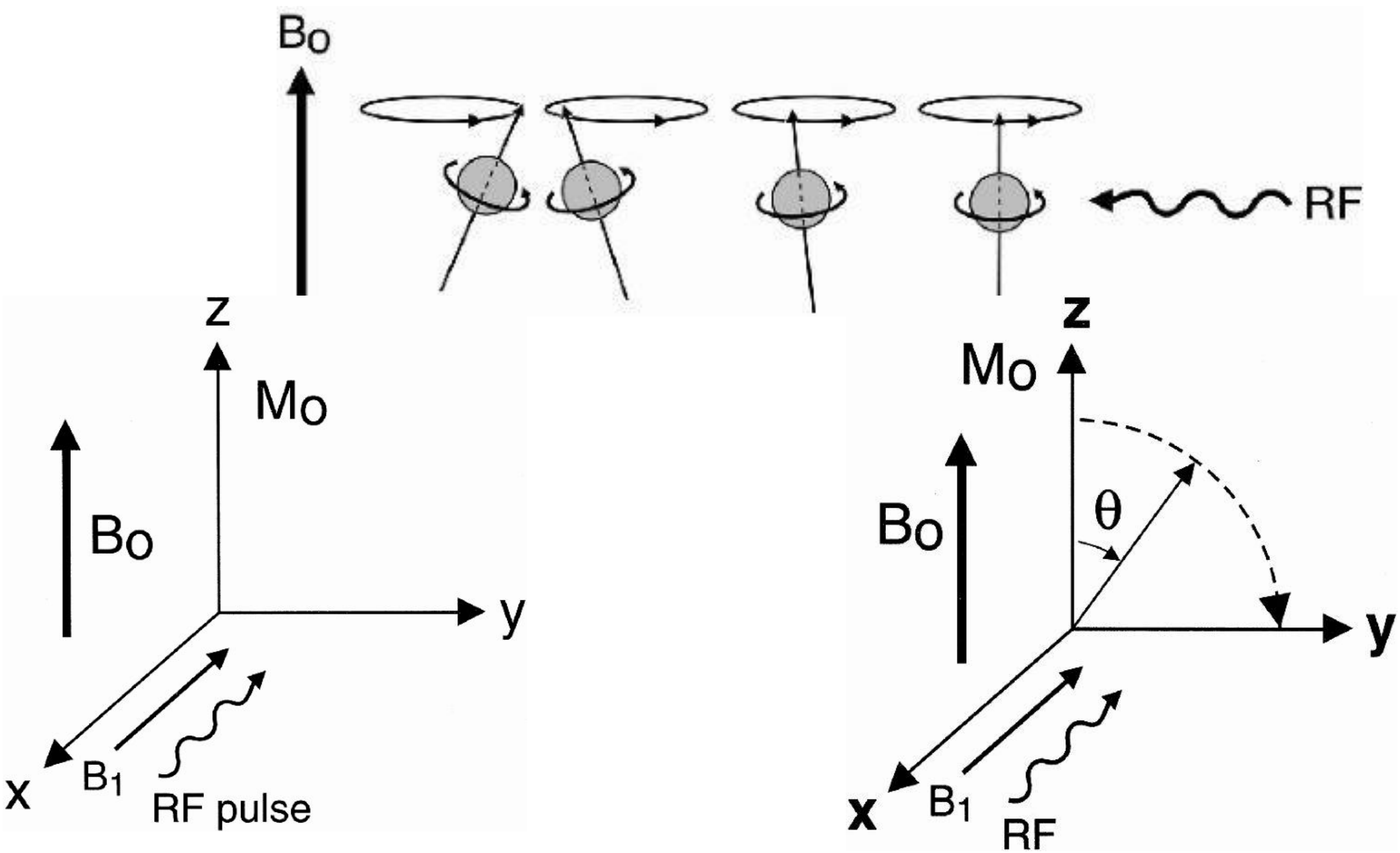
- B0 field is much larger than tissue net magnetization
  - Impossible to measure net magnetization in the z-direction
  - Need to look at component on x-y plane
  - Problem: x-y components cancel out
- Measured using pick-up coils



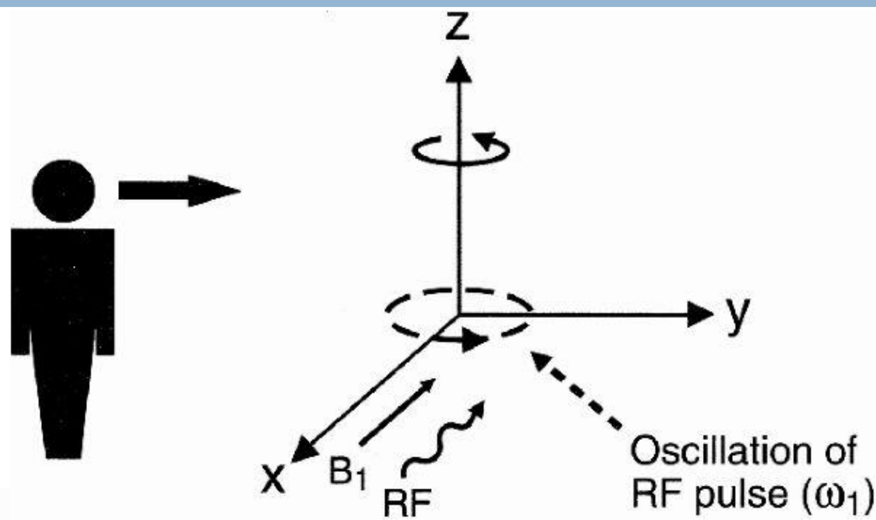


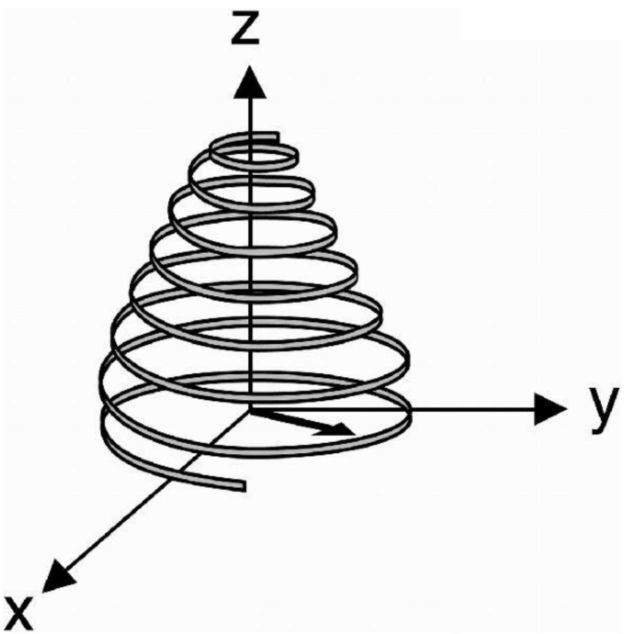
# RF Pulse

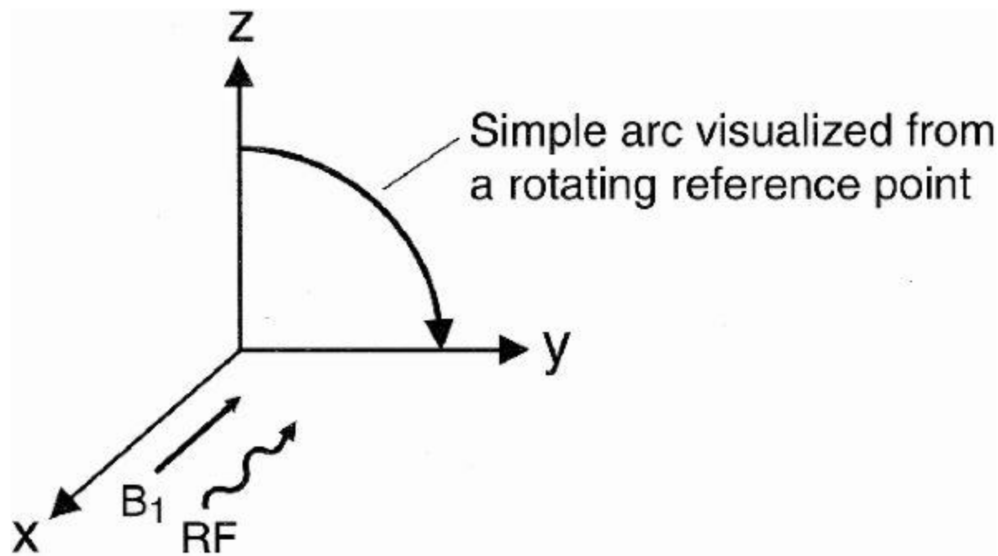
- Idea: Sending RF radiation at Larmor frequency to flip net magnetization to x-y plane



# Rotating Frame of Reference

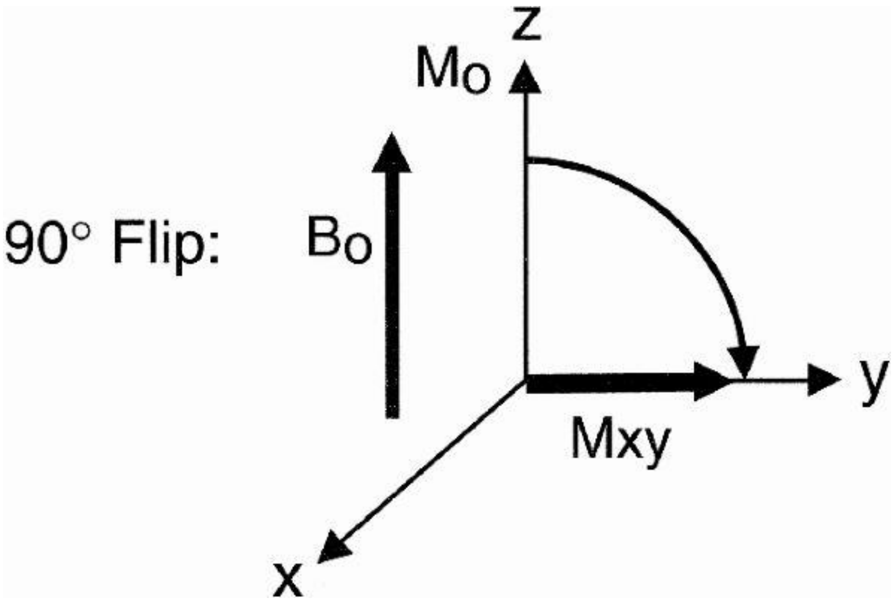


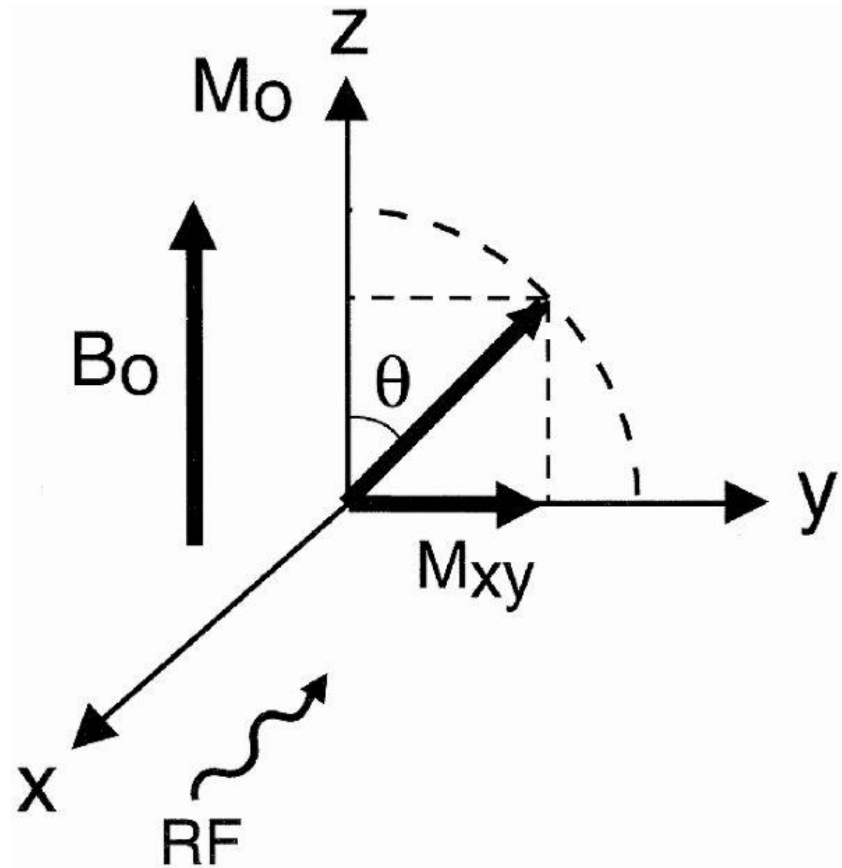



# Selection of RF Pulse Flip Angle

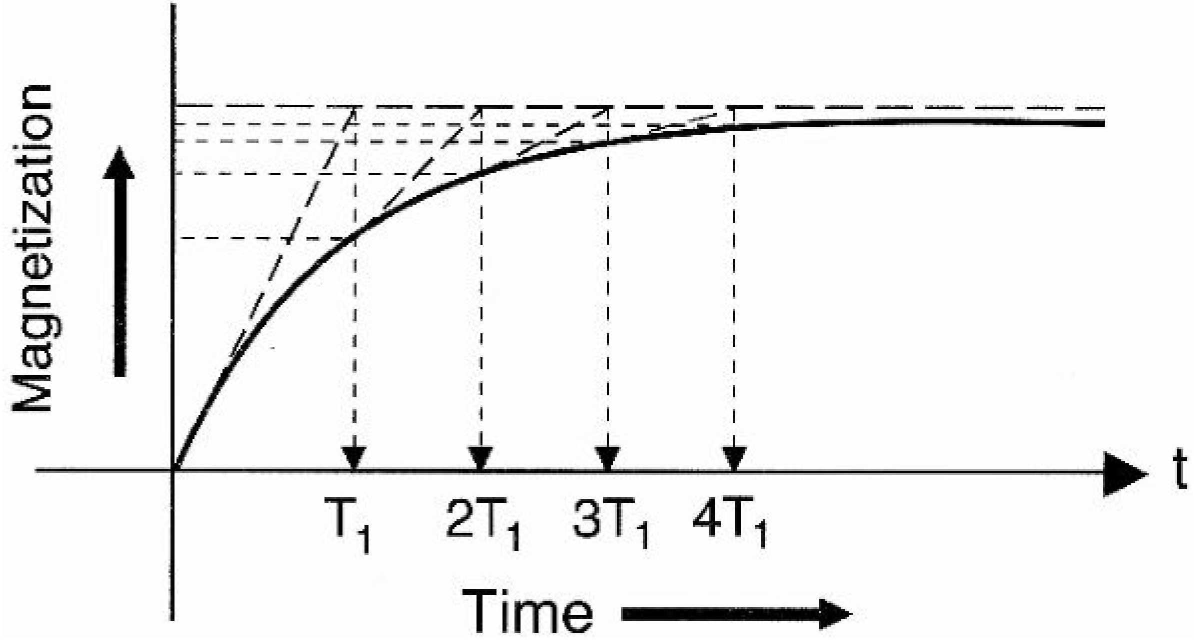
- 90° or 180° or partial flip



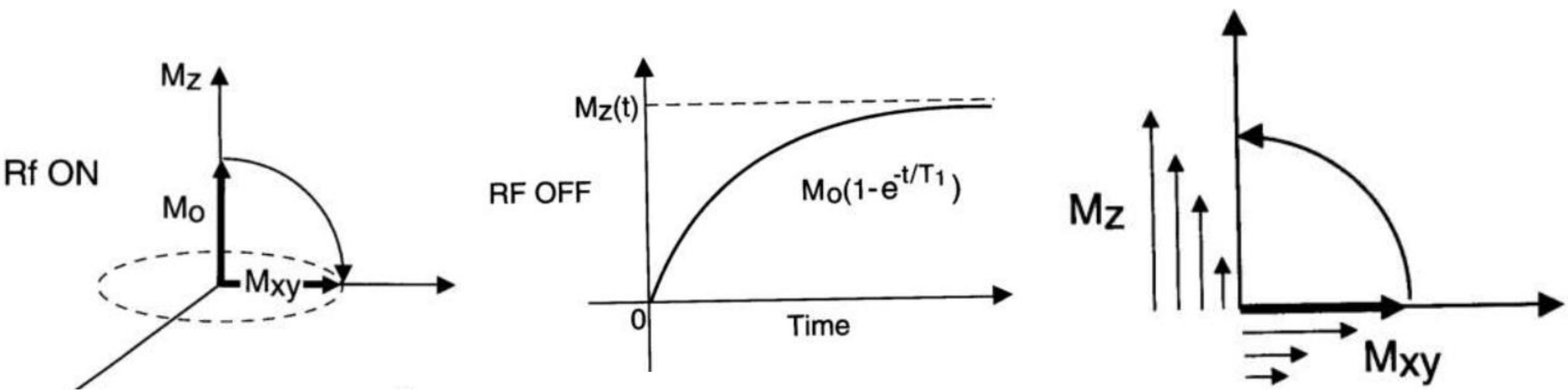
# Relaxation

- 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

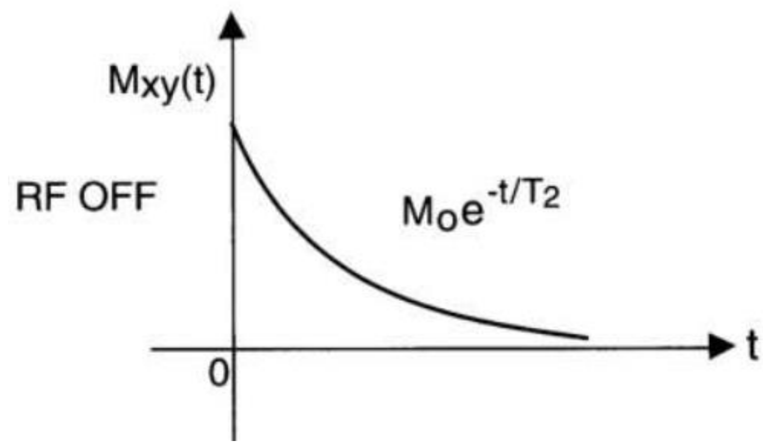
- T1 is called the longitudinal relaxation time because it refers to the time it takes for the spins to realign along the longitudinal (z) axis
- T1 is also called the spin-lattice relaxation time because it refers to the time it takes for the spins to give the energy they obtained from the RF pulse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

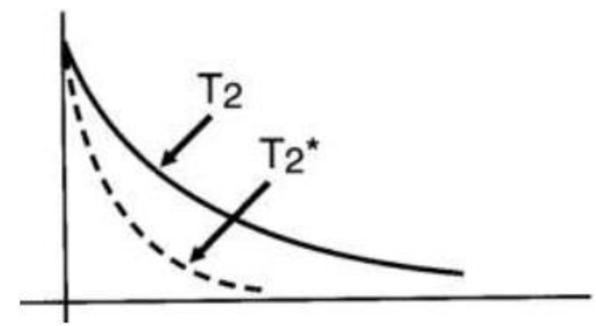


# T2 Relaxation

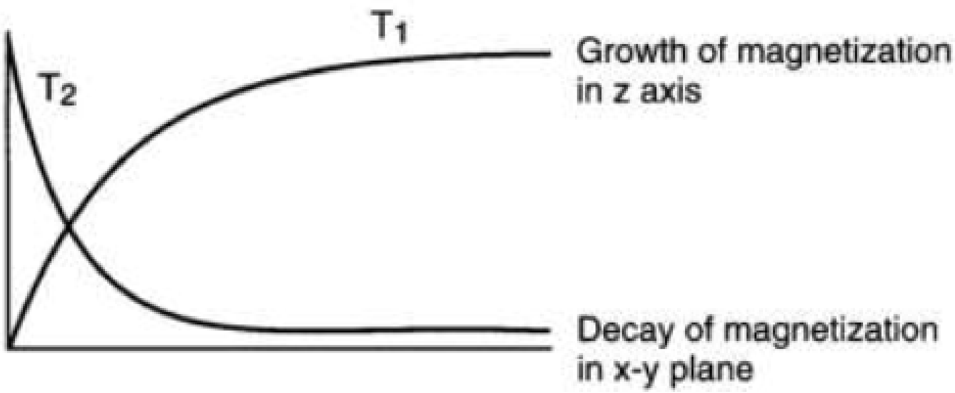
- Dephasing: after the 90° RF pulse is turned off, all spins are in phase; they are all lined up in the same direction and spinning at the same frequency $\omega_0$. There are two phenomena that will make the spins get out of phase: interactions between spins and external field inhomogeneities
- T2 Relaxation
  - Only spin-spin interactions
- T2* Relaxation
  - Both effects

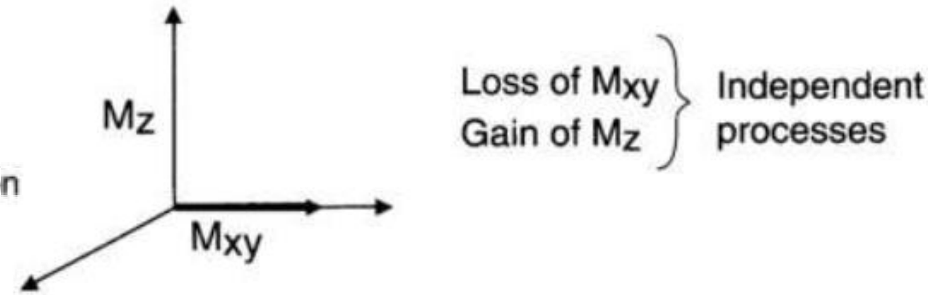
$$1/T2^* = 1/T2 + \gamma\Delta B$$

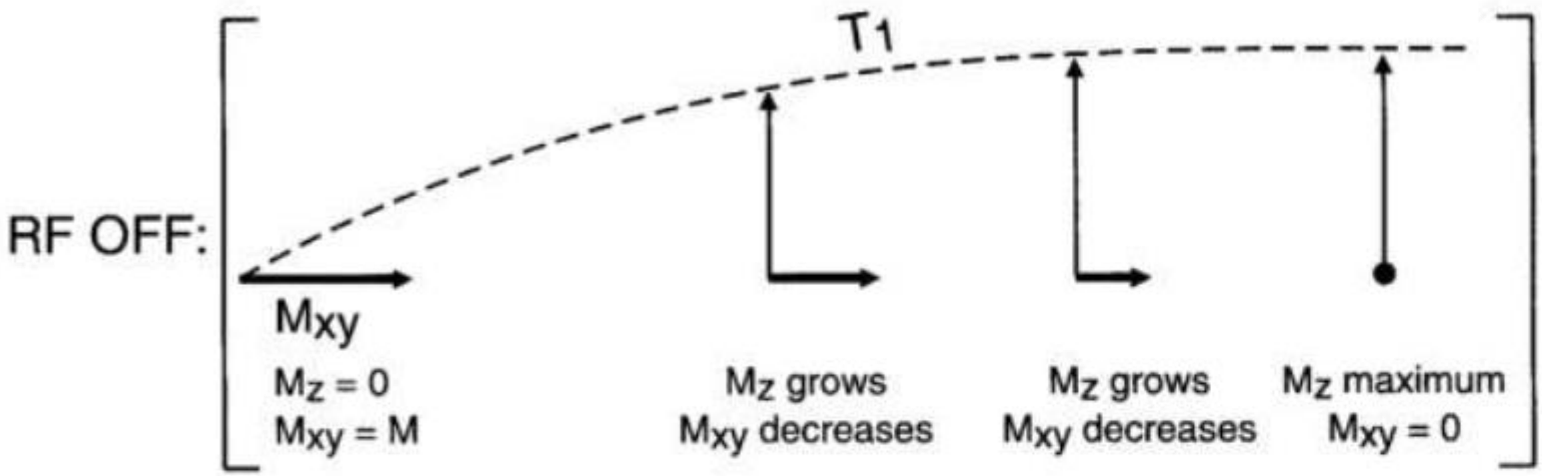


# Effect of Both T1 and T2 Relaxations



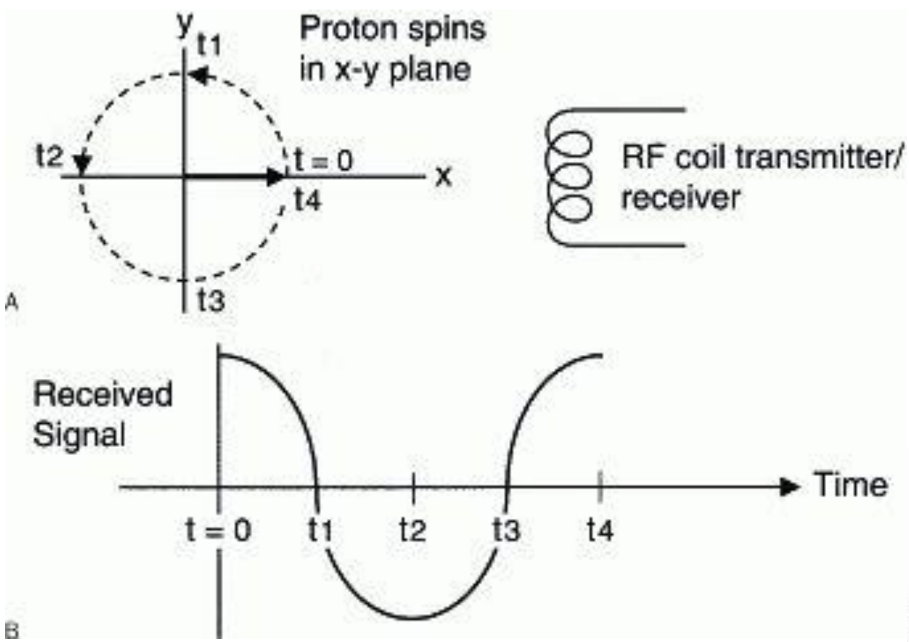



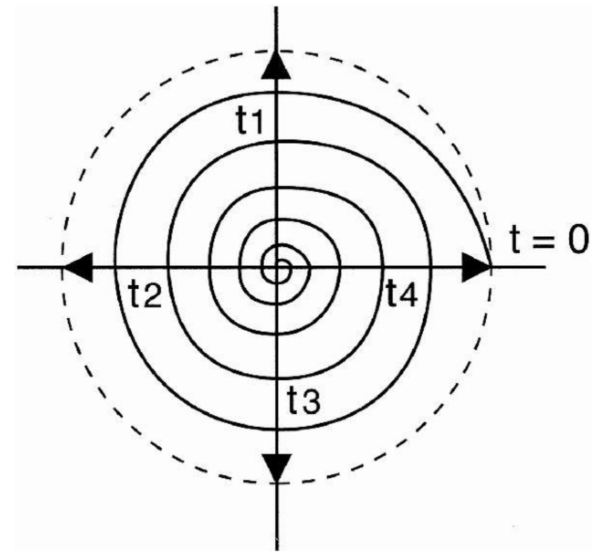



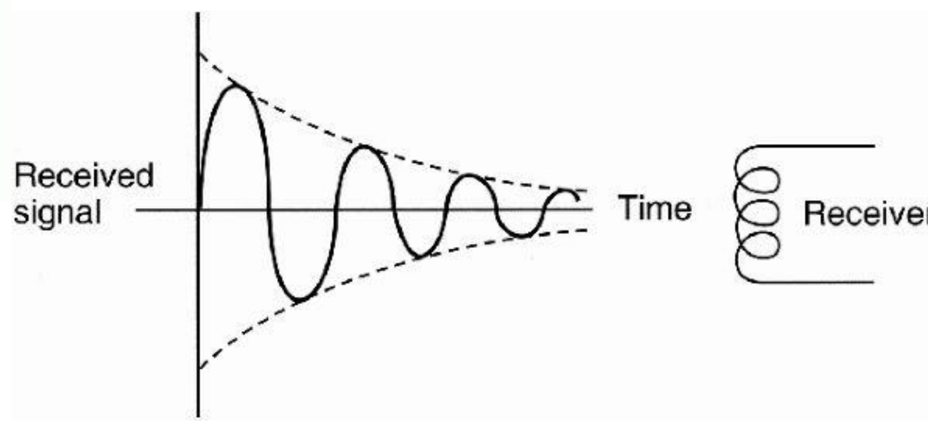
# Example Tissue Relaxation Times

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

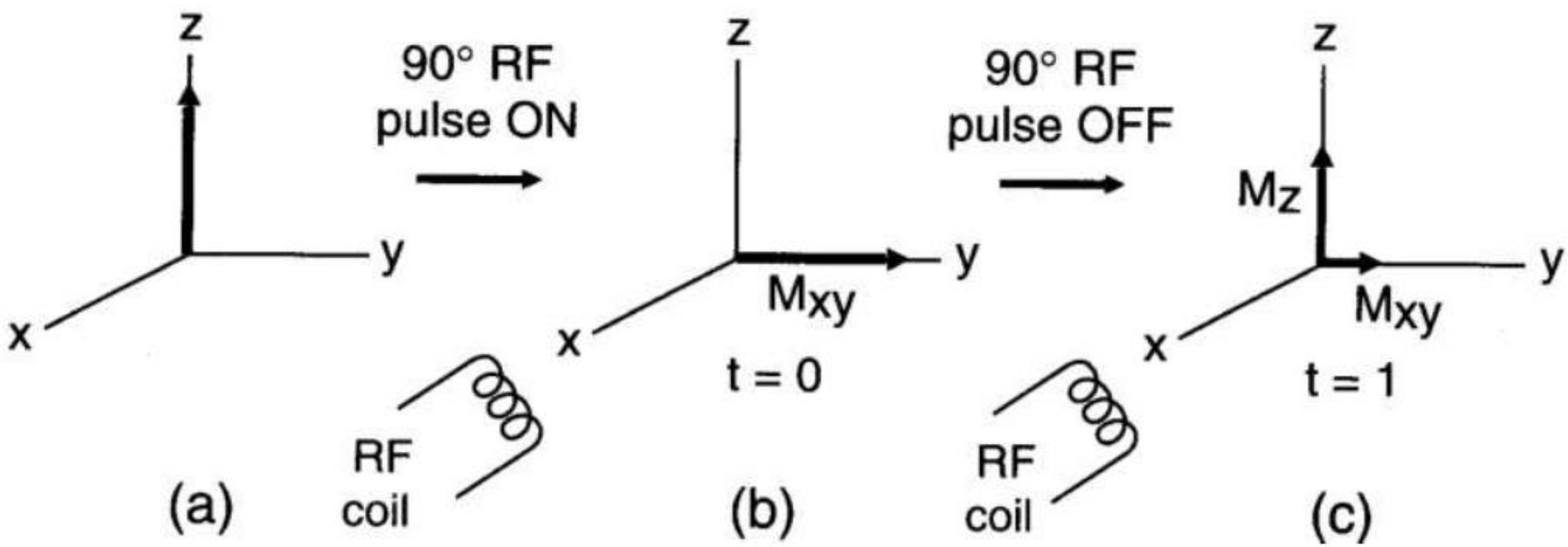
# Received Signal: Free Induction Decay (FID)

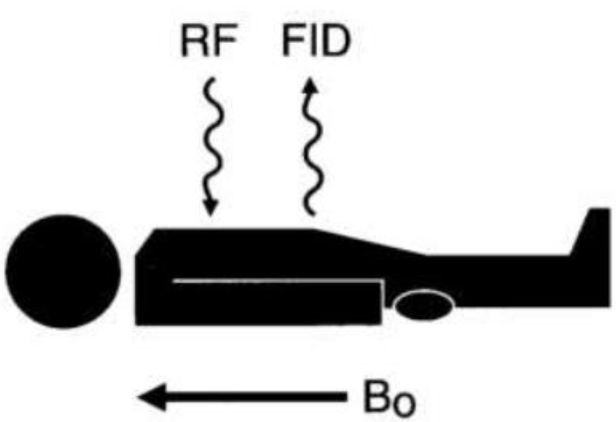


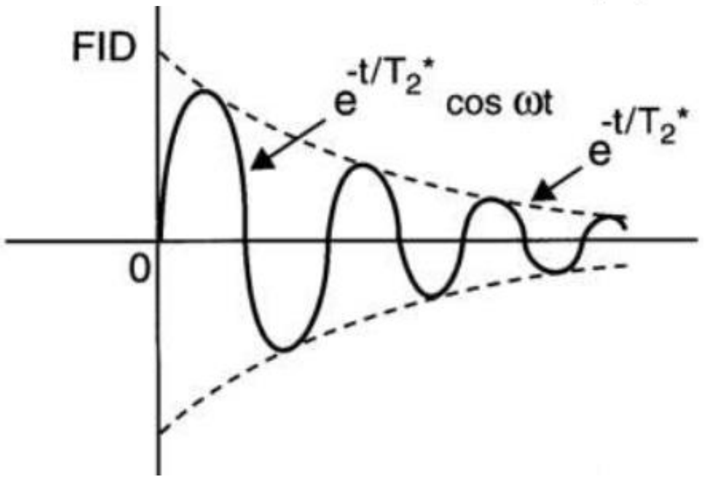



# Sequence of Events in MRI

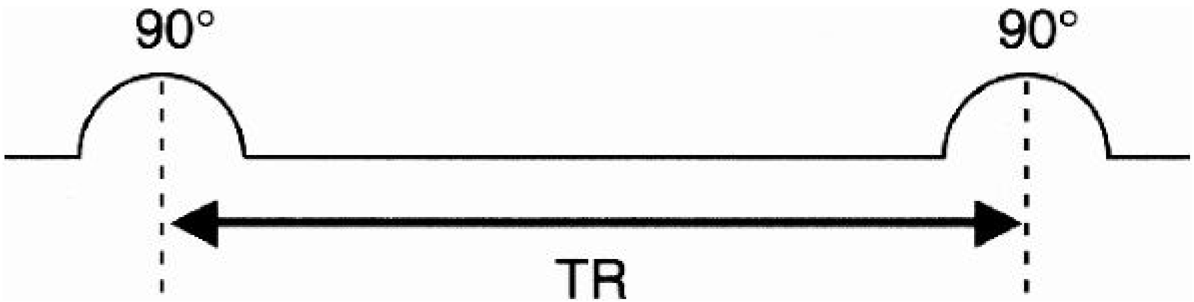


# Pulse Repetition Time (TR)

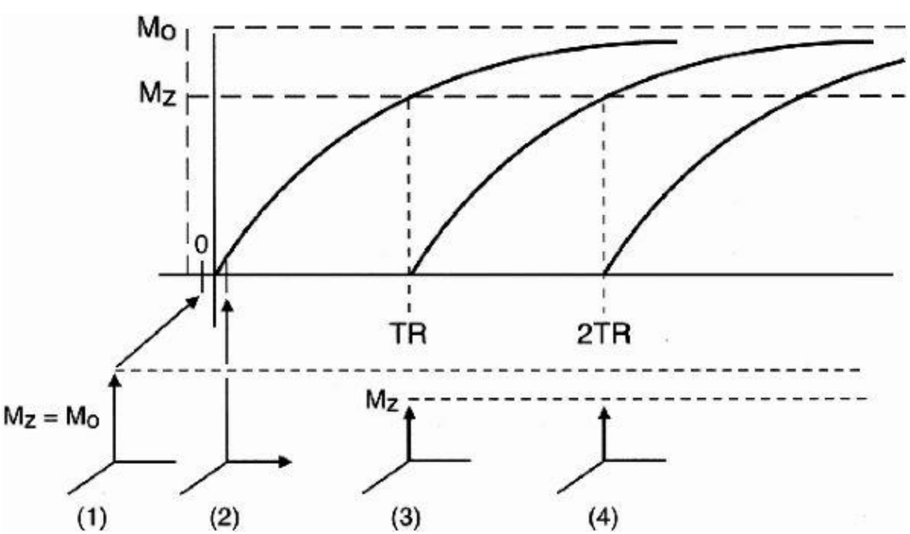
- Distance between successive RF pulses

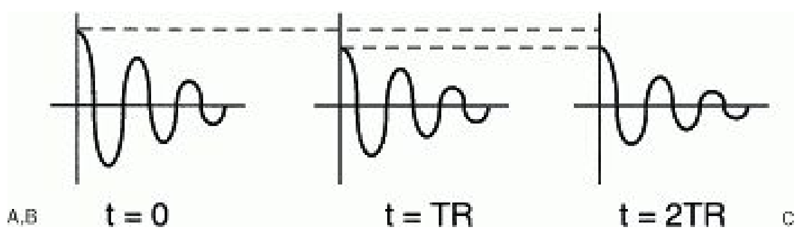



$M_z(t) = M_0 (1 - e^{-t/T1})$

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

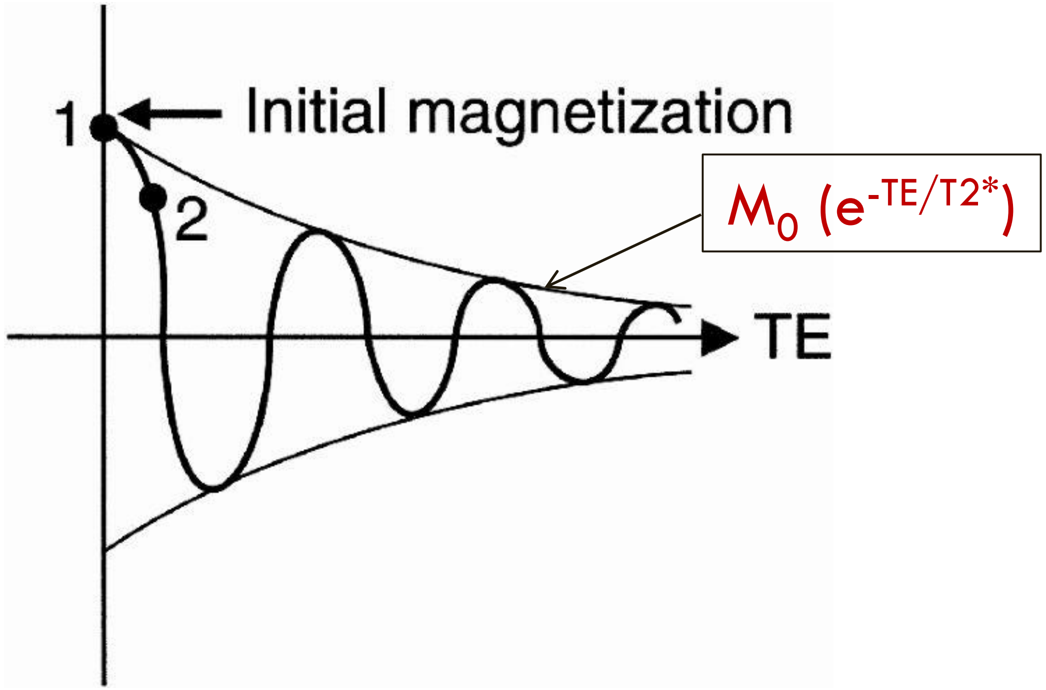
$S \propto N(H) (1 - e^{-TR/T1})$



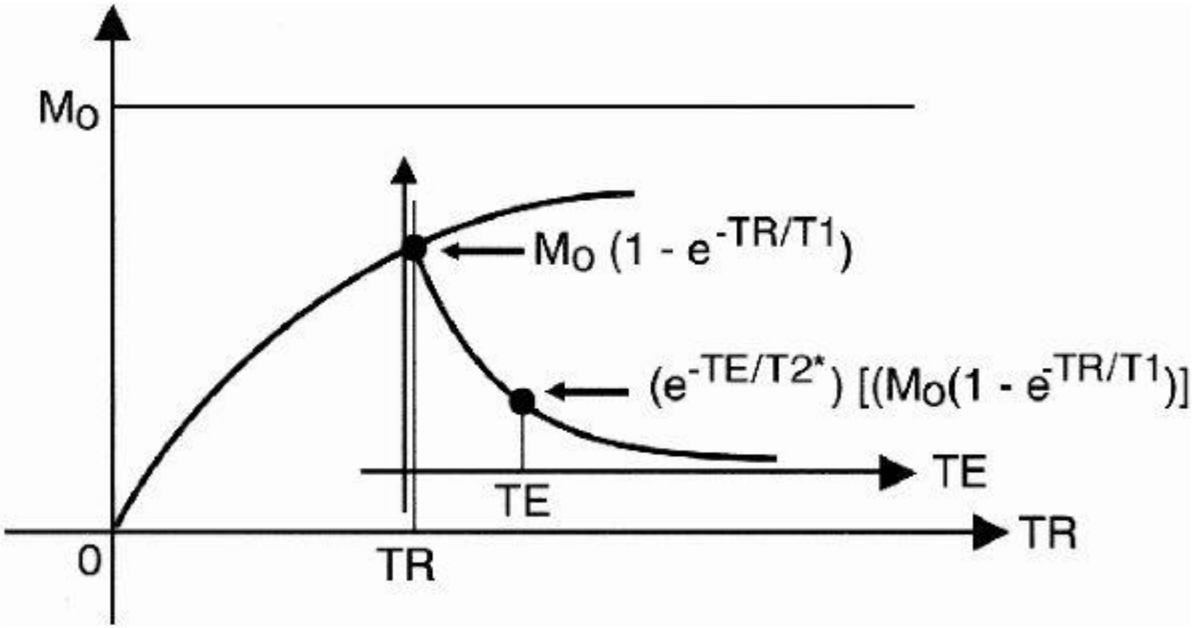
# Echo Time or Time to Echo (TE)

- 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

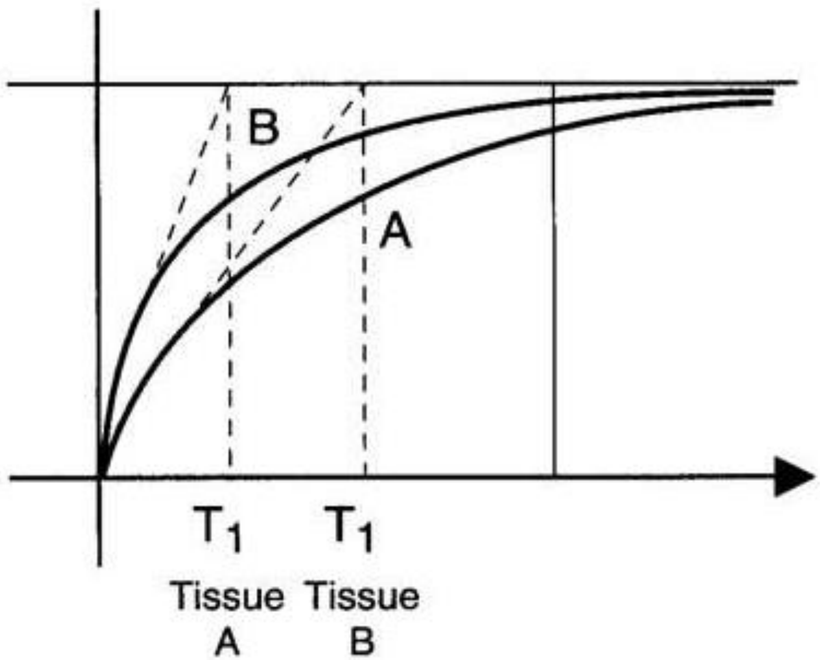
- Now we have to put the two curves together because both T1 recovery and T2 decay processes are occurring simultaneously

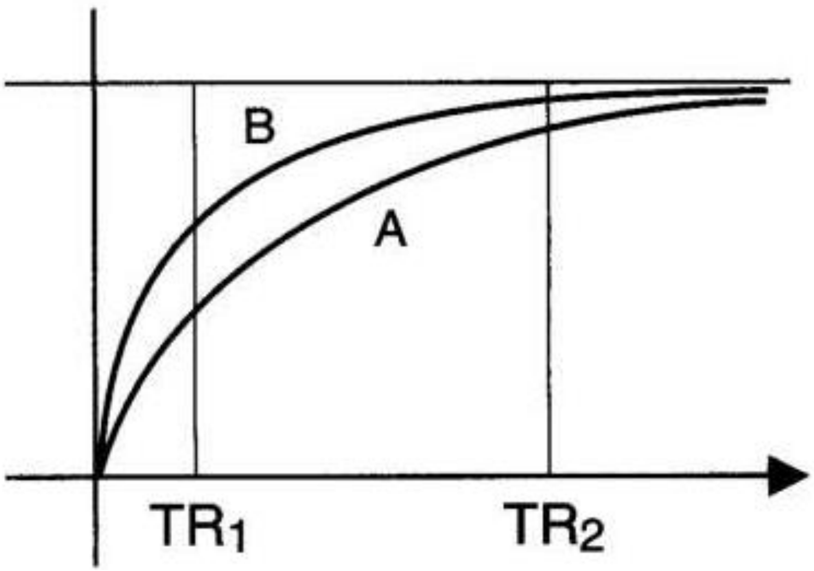



Signal Intensity = SI $\propto$ N(H)($e^{-TE/T2^*}$) (1 - $e^{-TR/T1}$)

# T1-Weighting

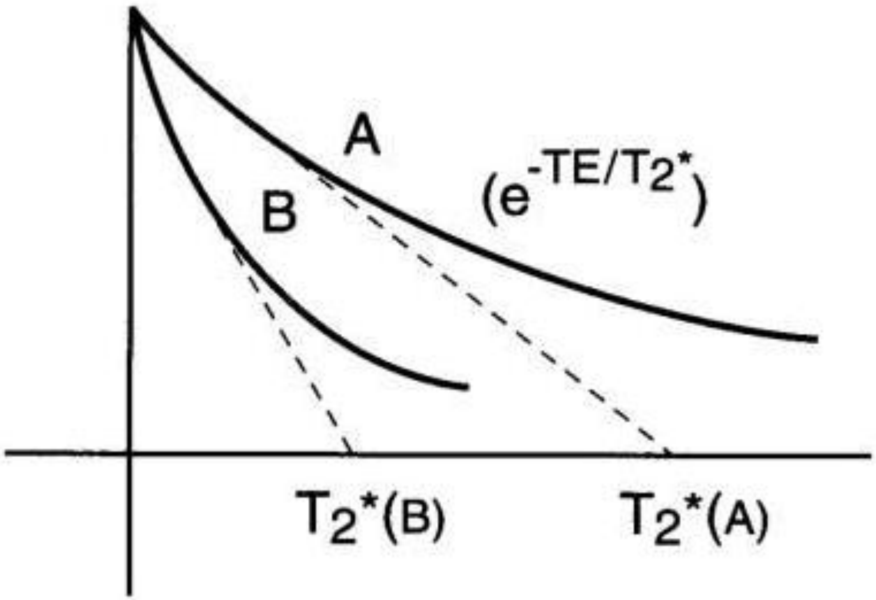
- Long TR reduces the T1 effect
- Short TR enhances the T1 contrast

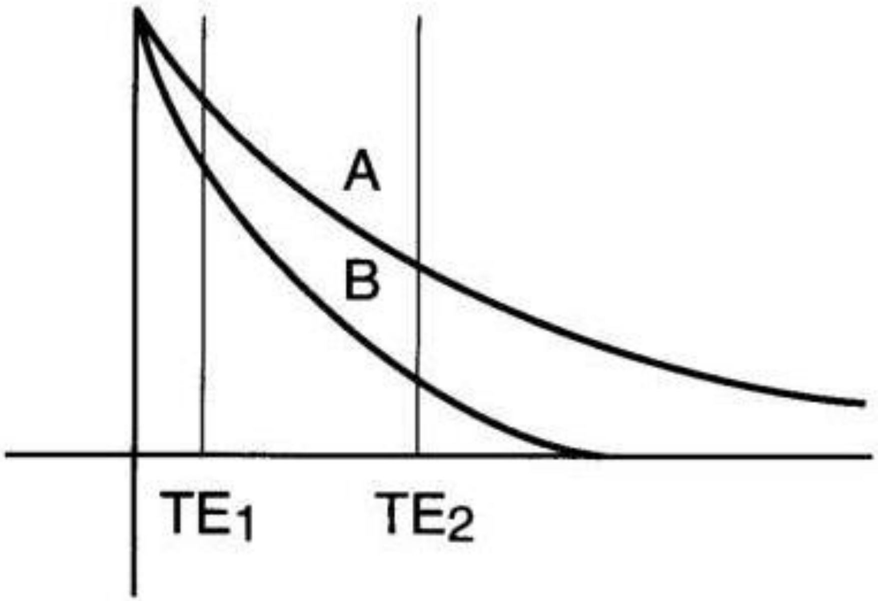


# T2-Weighting

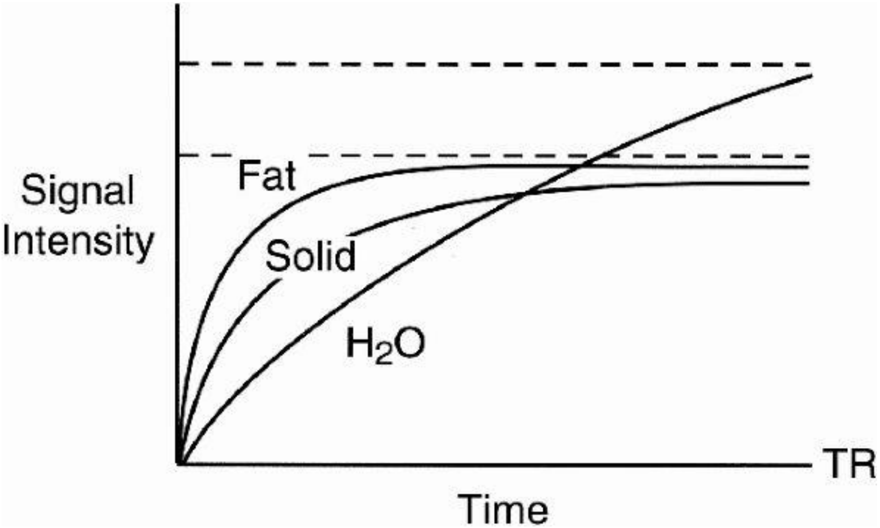
- Short TE reduces the T2* (T2) effect
- Long TE enhances the T2* (T2) effect

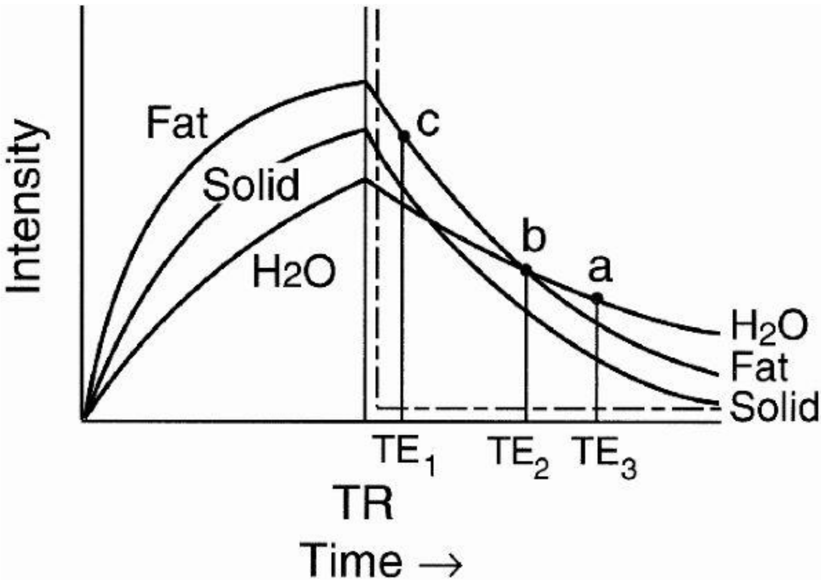


# Tissue Contrast: Clinical Applications

- 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

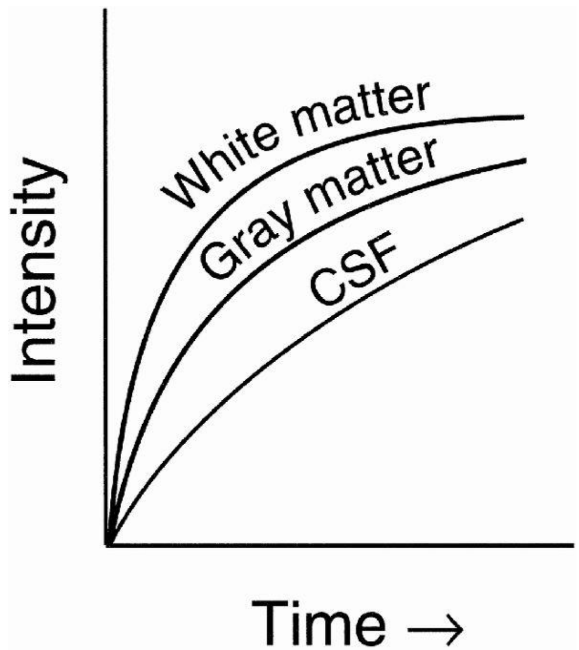
| | **Long T1 (low SI)** | **Intermediate** | **Short T1 (high SI)** |
|---|---|---|---|
| **Long T2 (high SI)** | Water/CSF<br>Pathology<br>Edema | | d (EC metHgb) |
| **Intermediate** | | Muscle<br>GM<br>a (oxyHgb)<br>WM | |
| **Short T2 (low SI)** | Air<br>Cortical bone<br>Heavy $Ca^{++}$<br>b (deoxyHgb)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

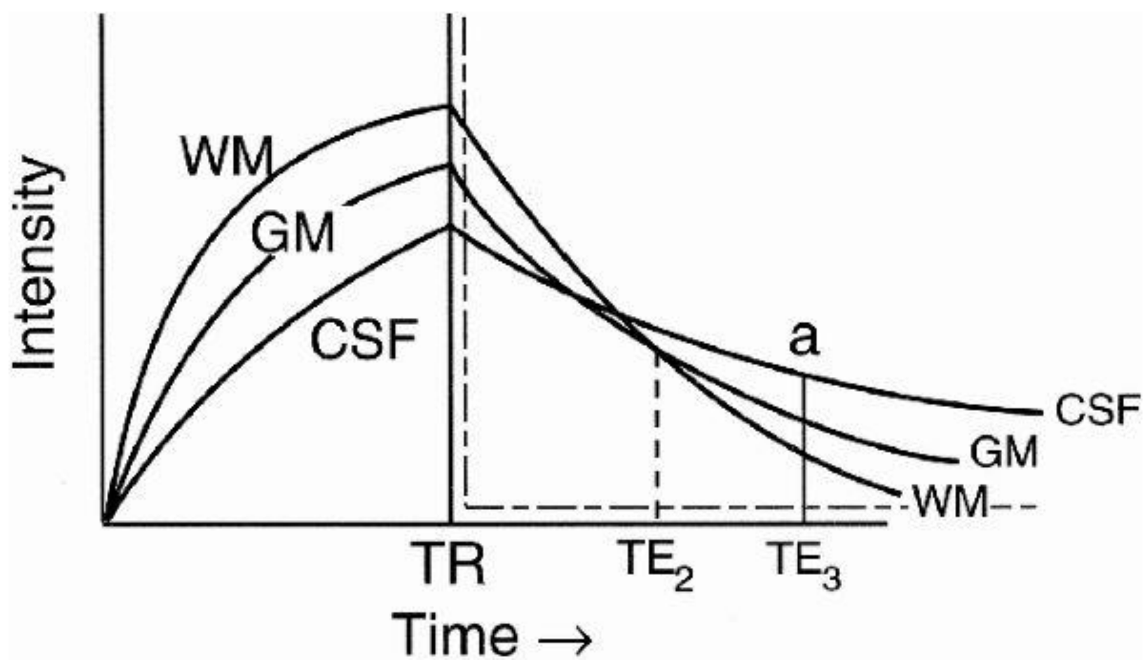
a-d represent breakdown products of hemoglobin (a, oxyhemoglobin; b, deoxyhemoglobin; c, intracellular methemoglobin; d, extracellular methemoglobin; e, hemosiderin). GM, gray matter; WM, white matter; SI, signal intensity; Hgb, hemoglobin; IC, intracellular; EC, extracellular.

# Example: Brain Imaging

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

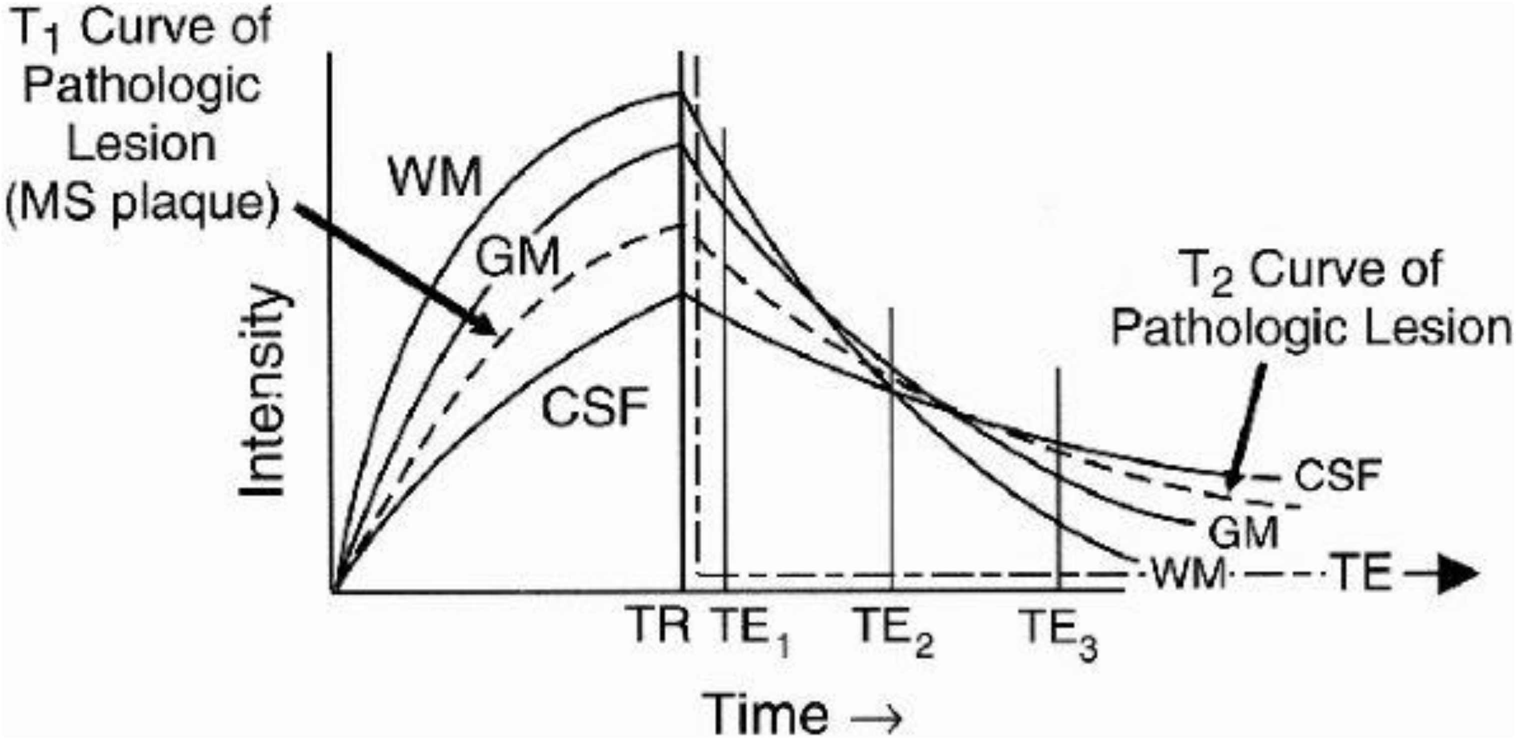
| | $T_1$ (msec) | $T_2$ (msec) | N(H) |
|---|---|---|---|
| White matter | 510 | 67 | 0.61 |
| Gray matter | 760 | 77 | 0.69 |
| Edema | 900 | 126 | 0.86 |
| CSF | 2650 | 180 | 1.00 |



# Example: Brain Imaging

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

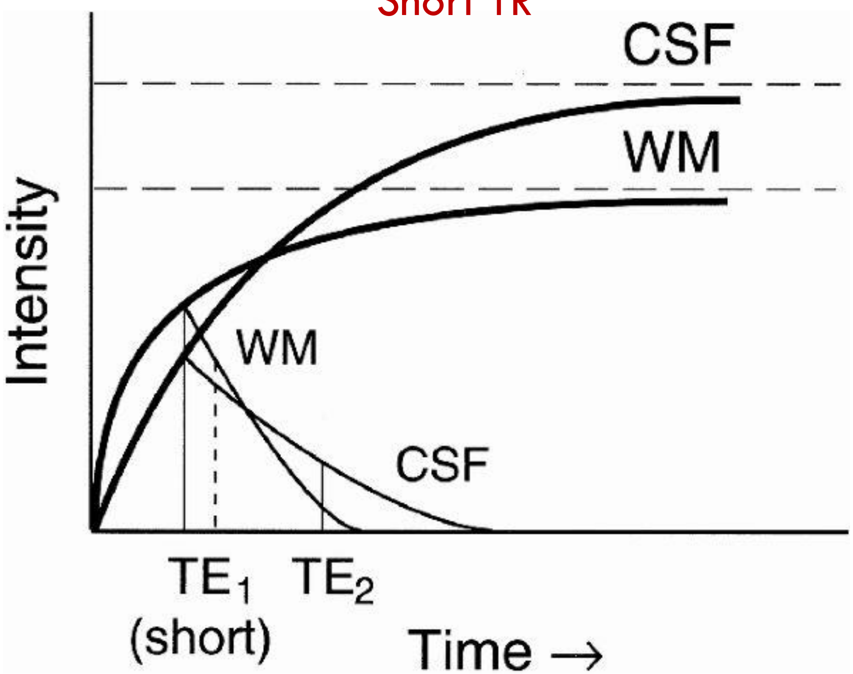


# Tissue Contrast: T1W, T2W and PDW

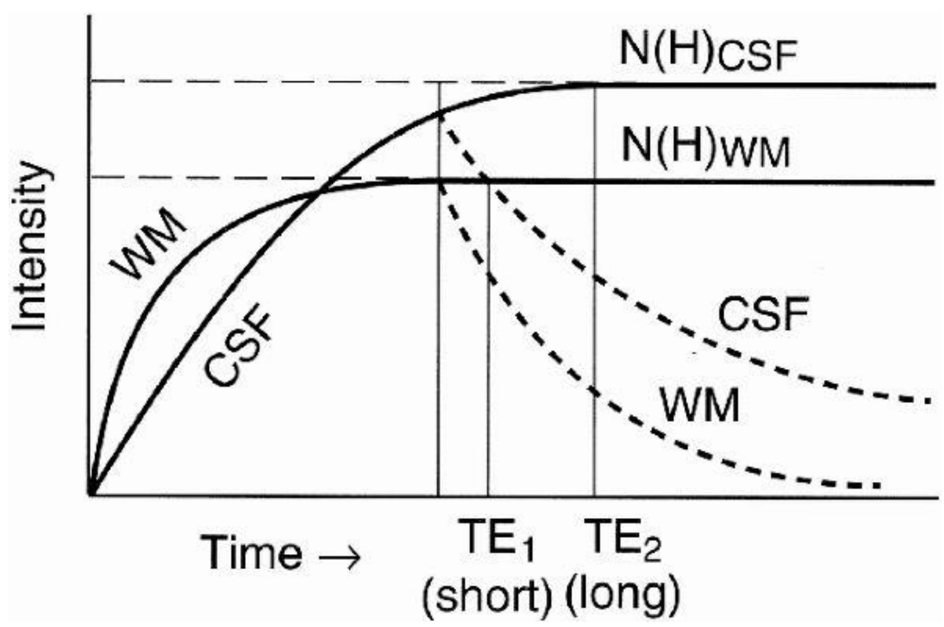
- Consider Cases of:
  - Short/Long TR
  - Short/Long TE

| | $T_1$ (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



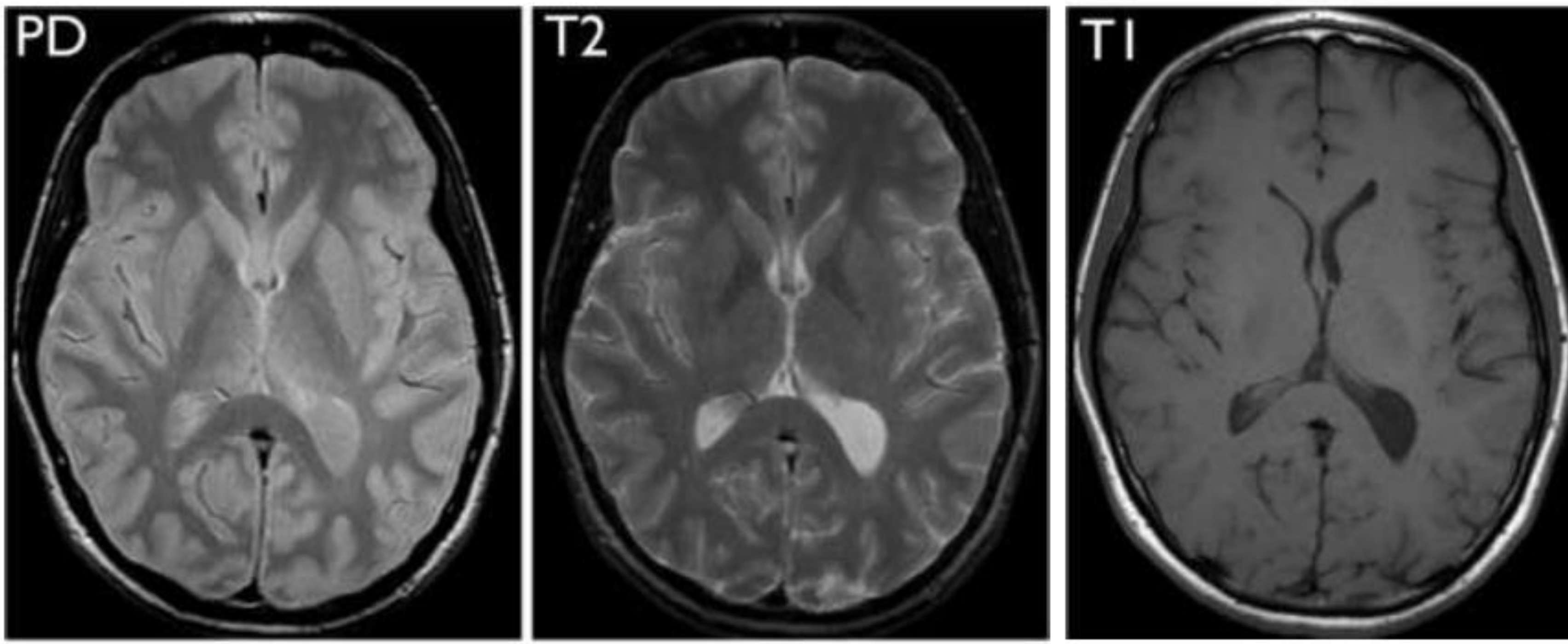
# Tissue Contrast Summary

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

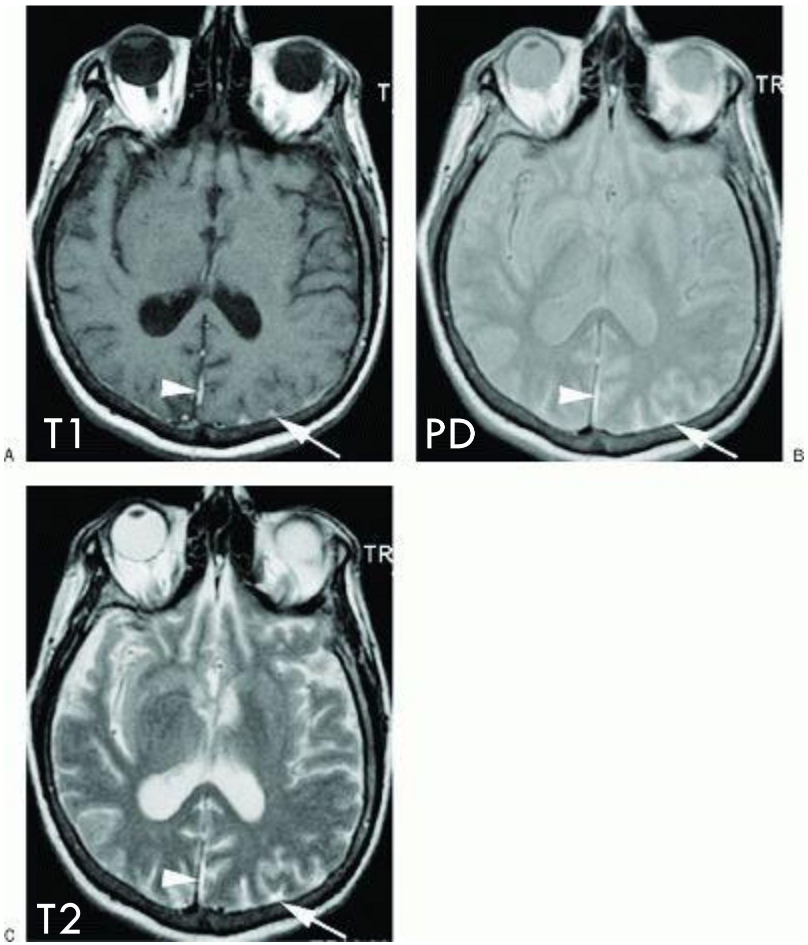
| | Short TE | Long TE |
|---|---|---|
| short TR | T1W | mixed |
| long TR | PDW | T2W |

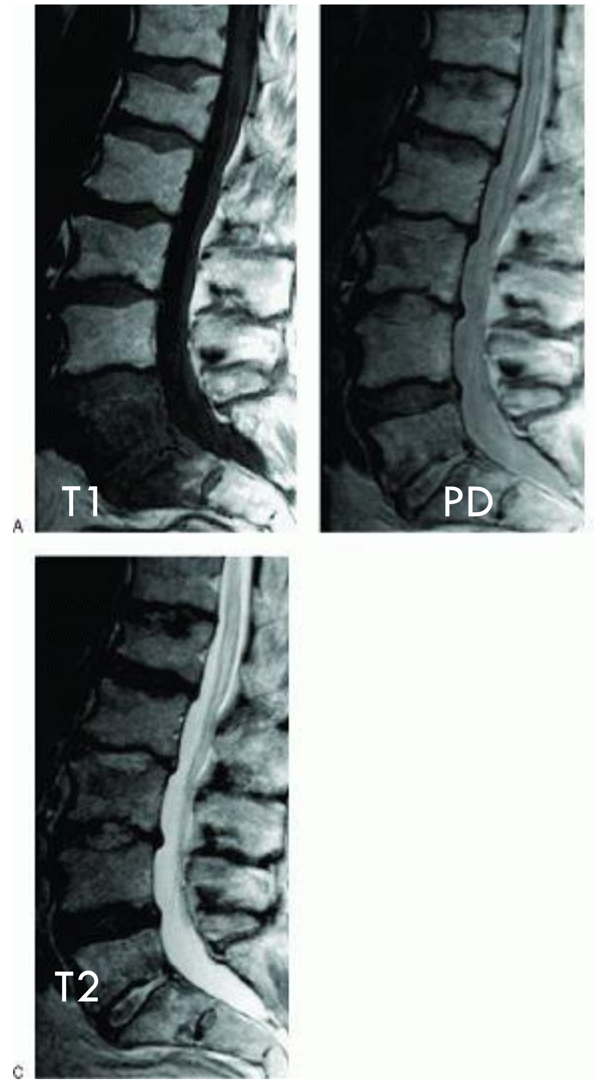
# Tissue Contrast Examples

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

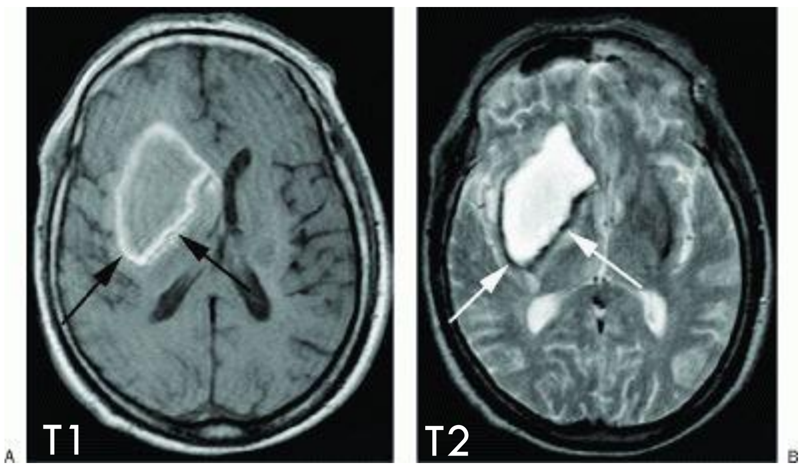


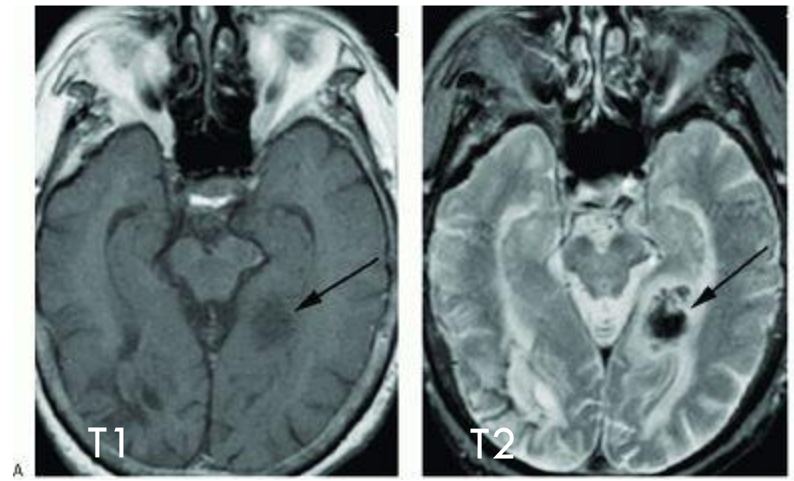
# Tissue Contrast Examples

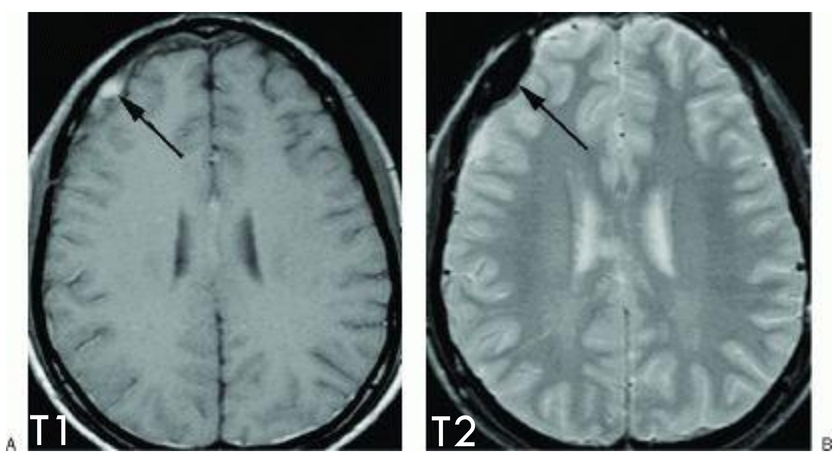


# Tissue Contrast Examples

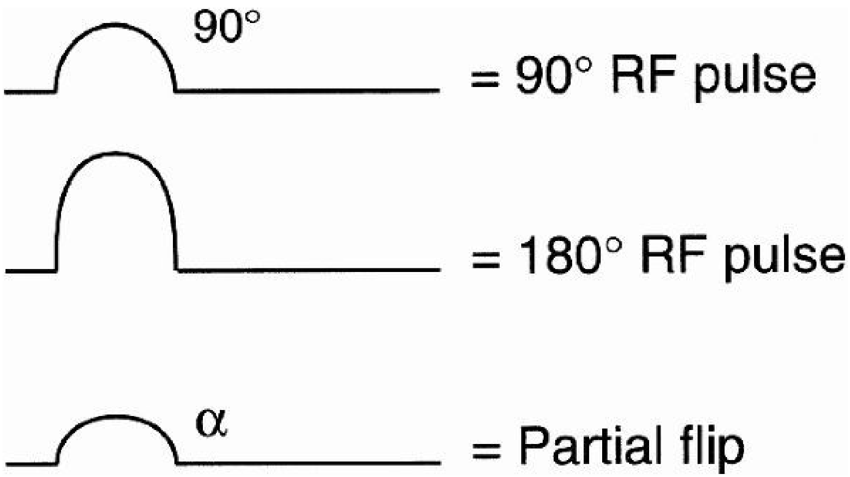


# Pulse Sequences: Saturation, Saturation Recovery and Inversion Recovery

- 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

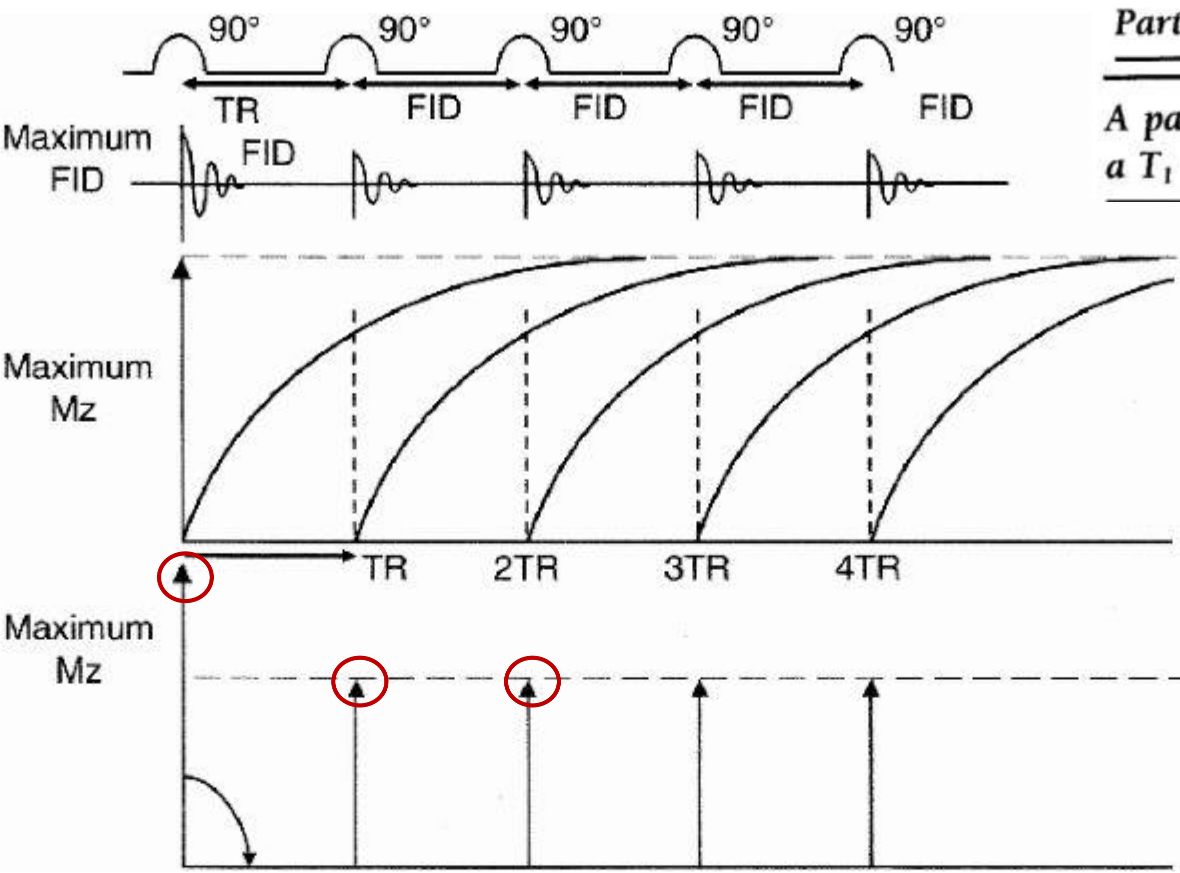


# Saturation

- 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.



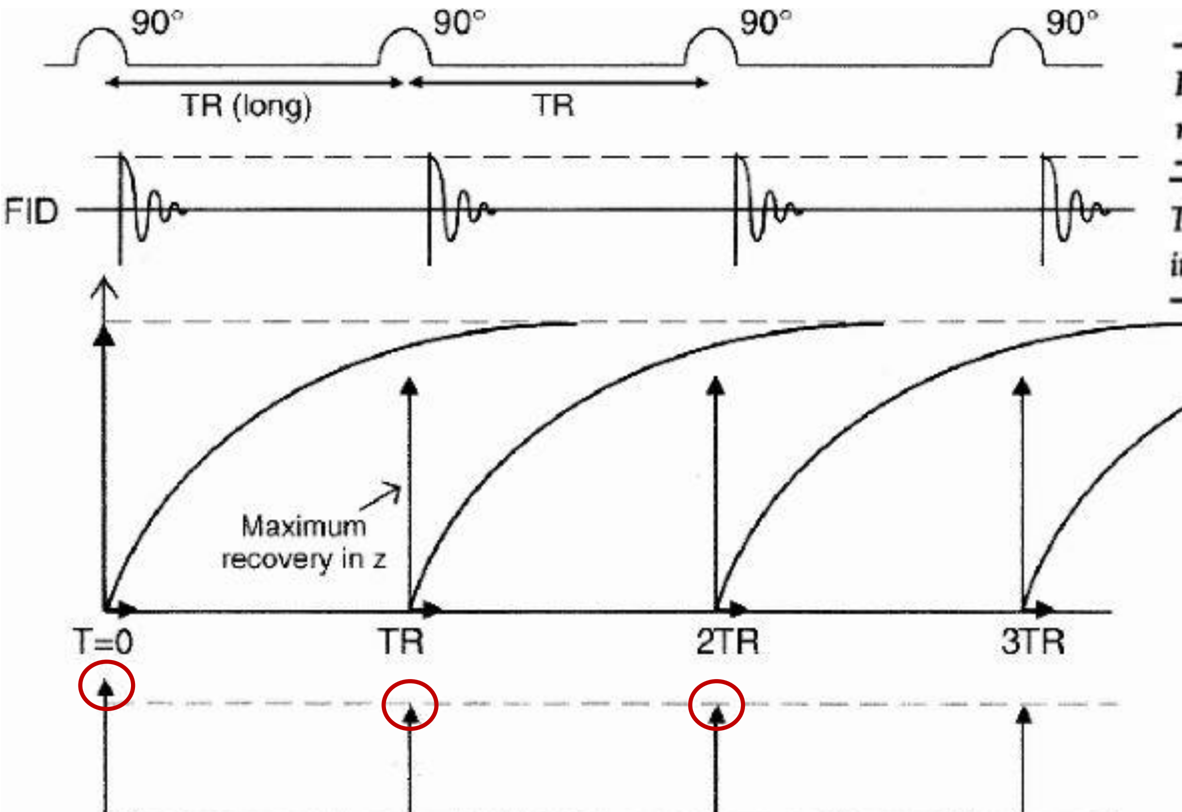

*Partial saturation: TR is short, TE is minimal.*

*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)



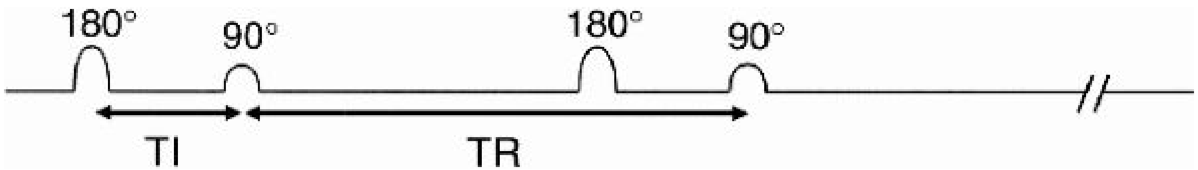

*In saturation recovery, TR is long and TE is minimal.*

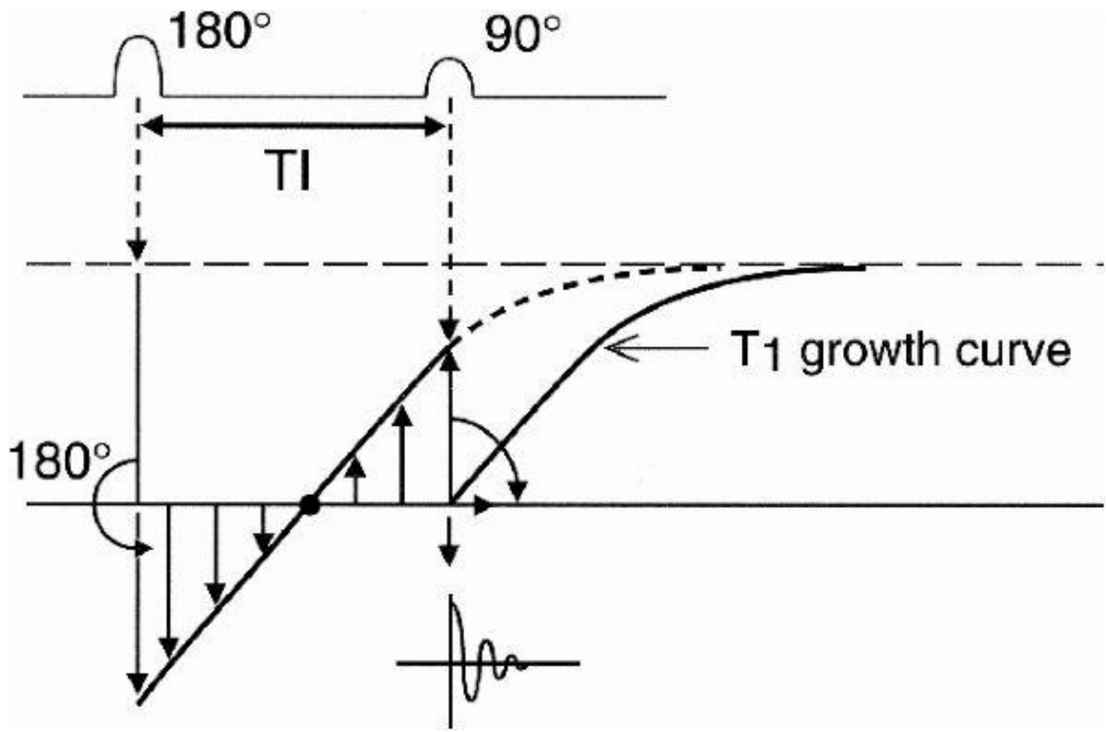
*The saturation recovery pulse sequence results in a proton density weighted image.*

Longitudinal Magnetization almost completely recovered

# Inversion Recovery Pulse Sequence

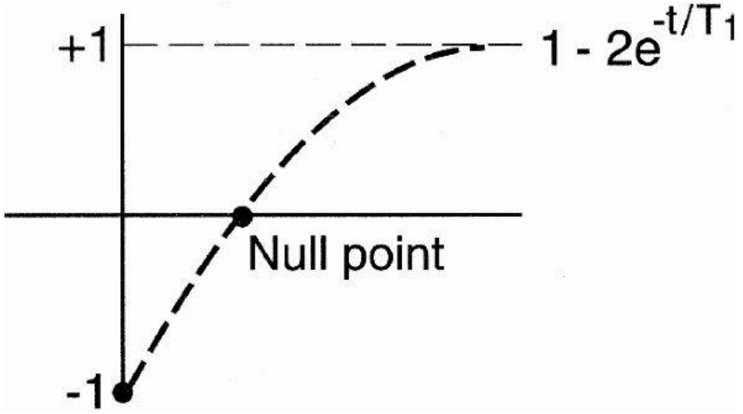
- first apply a 180° RF pulse. Next, we wait a period of time (the inversion time TI) and apply a 90° RF pulse

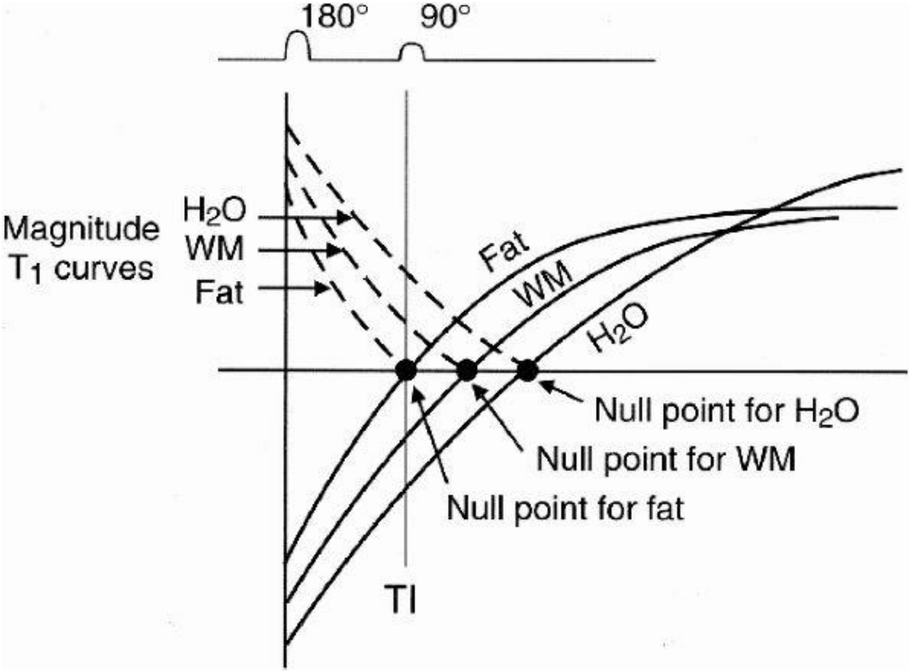


# Inversion Recovery: Null Point

- 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 **T**I **I**nversion **R**ecovery




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

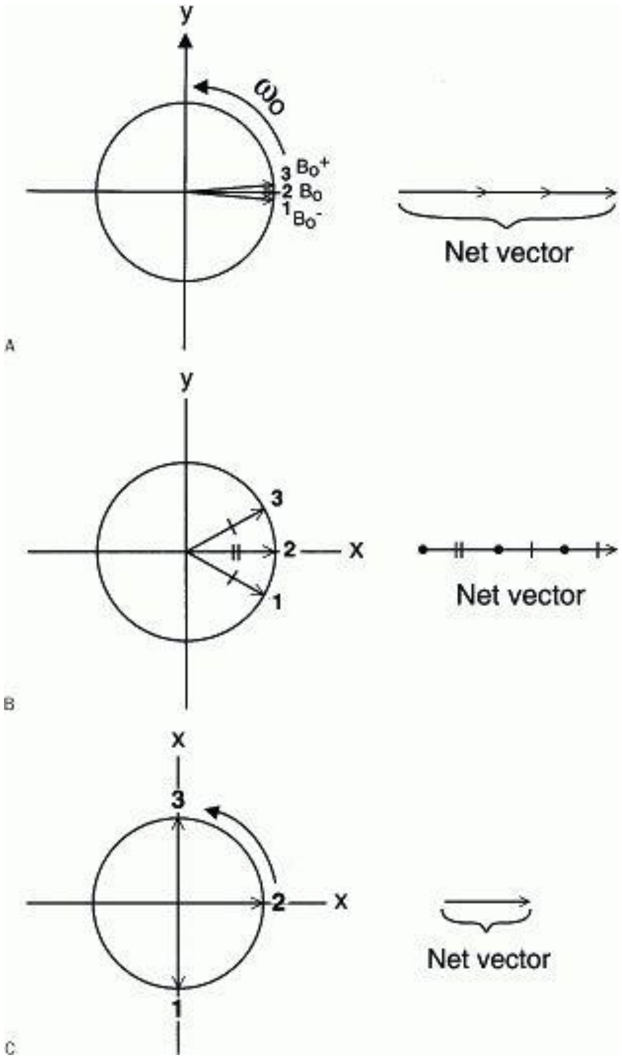
$$TI\,(\text{null}) = 0.693 \times T1$$

# Pulse Sequences: Spin Echo

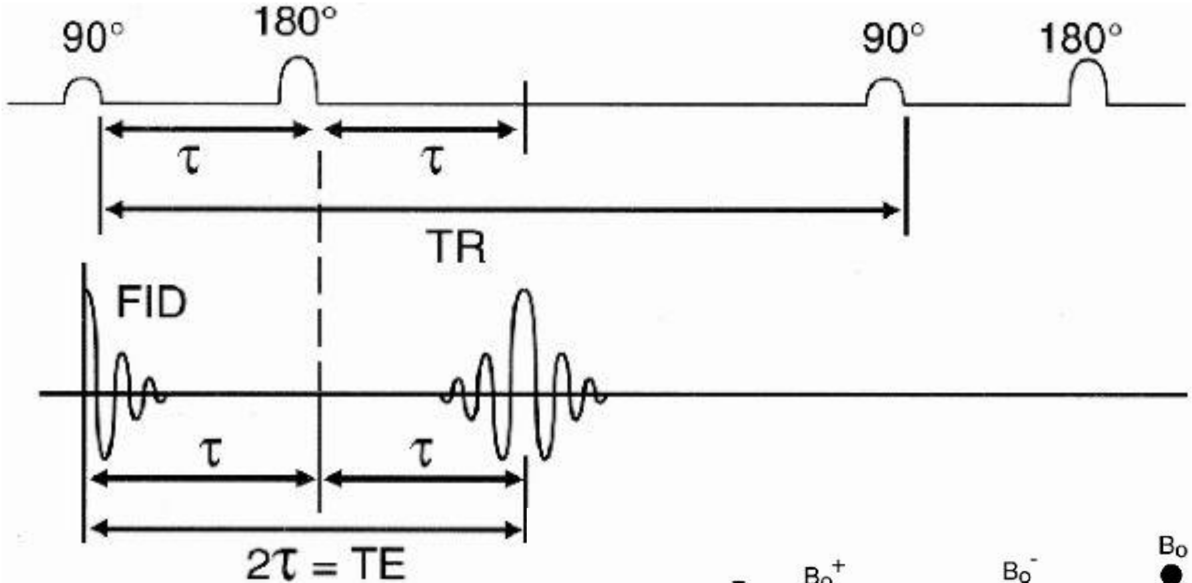
- Dephasing problem causes
  - Spin-spin interactions (inherent)
  - External magnetic field inhomogeneity

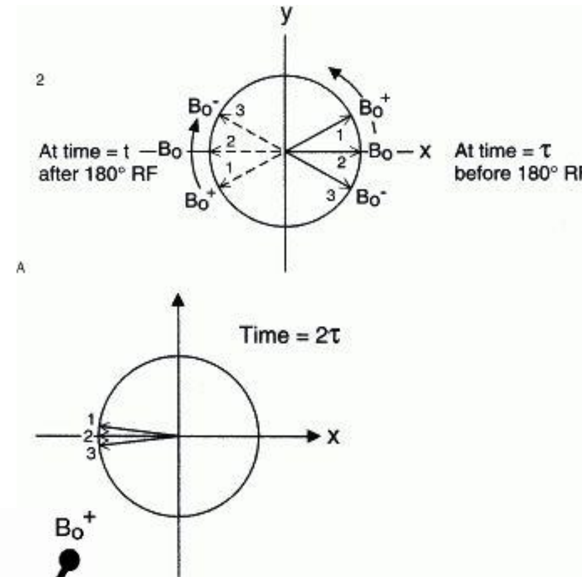
$$1/T2^* = 1/T2 + \gamma \Delta B$$

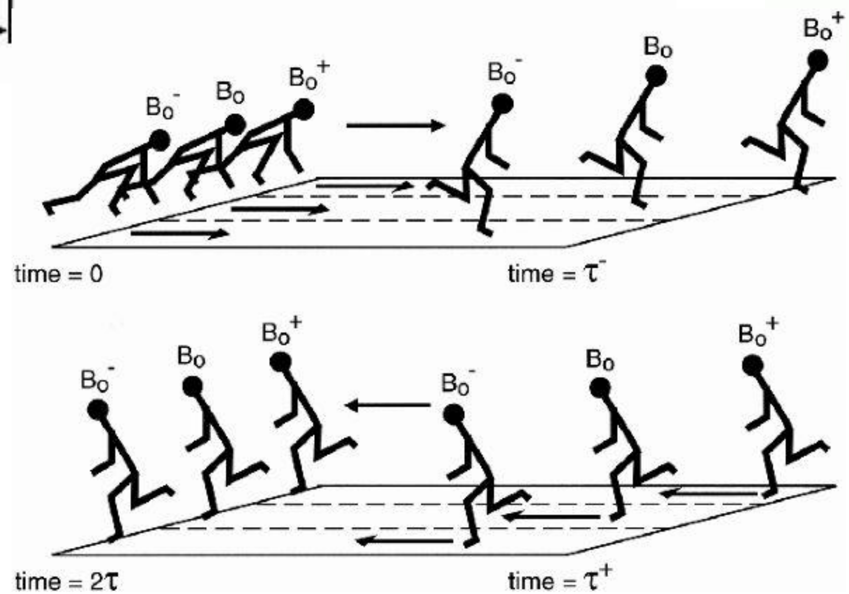
- Spin echo sequence: only T2



# Spin Echo Pulse Sequence

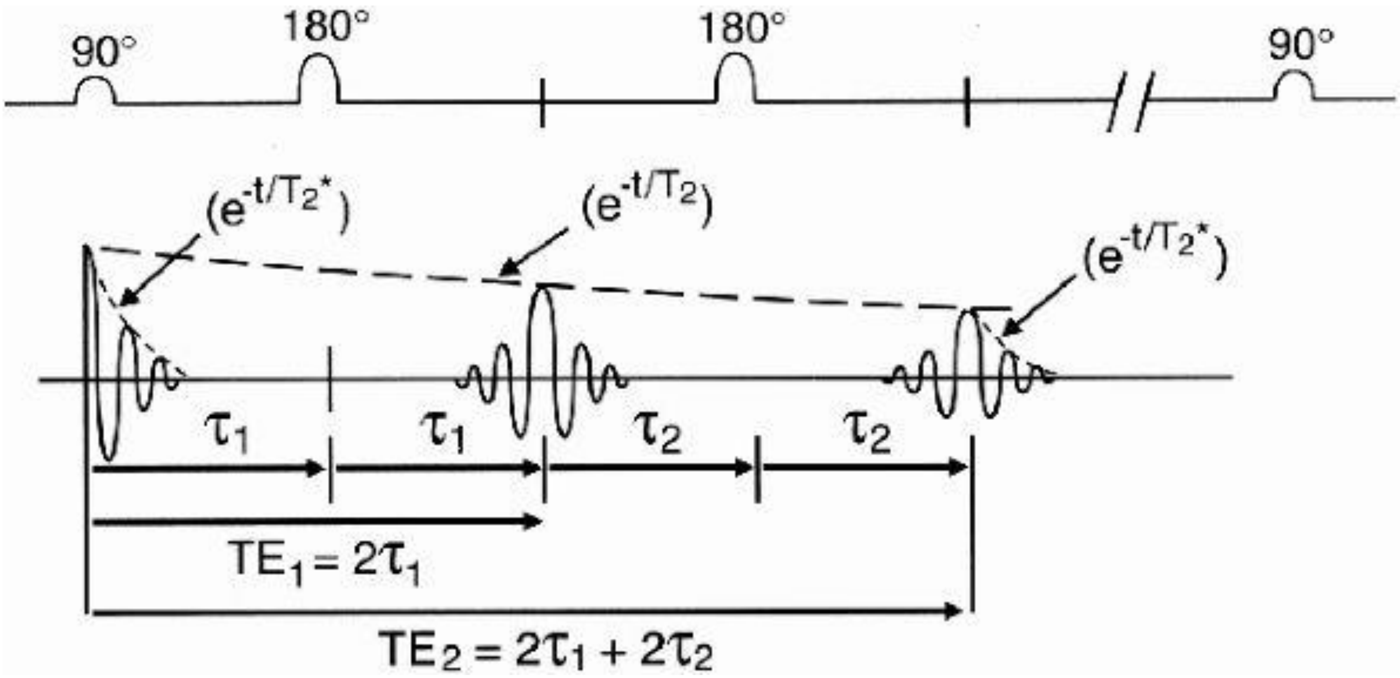


# Multi-Echo Spin Echo Pulse Sequence

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



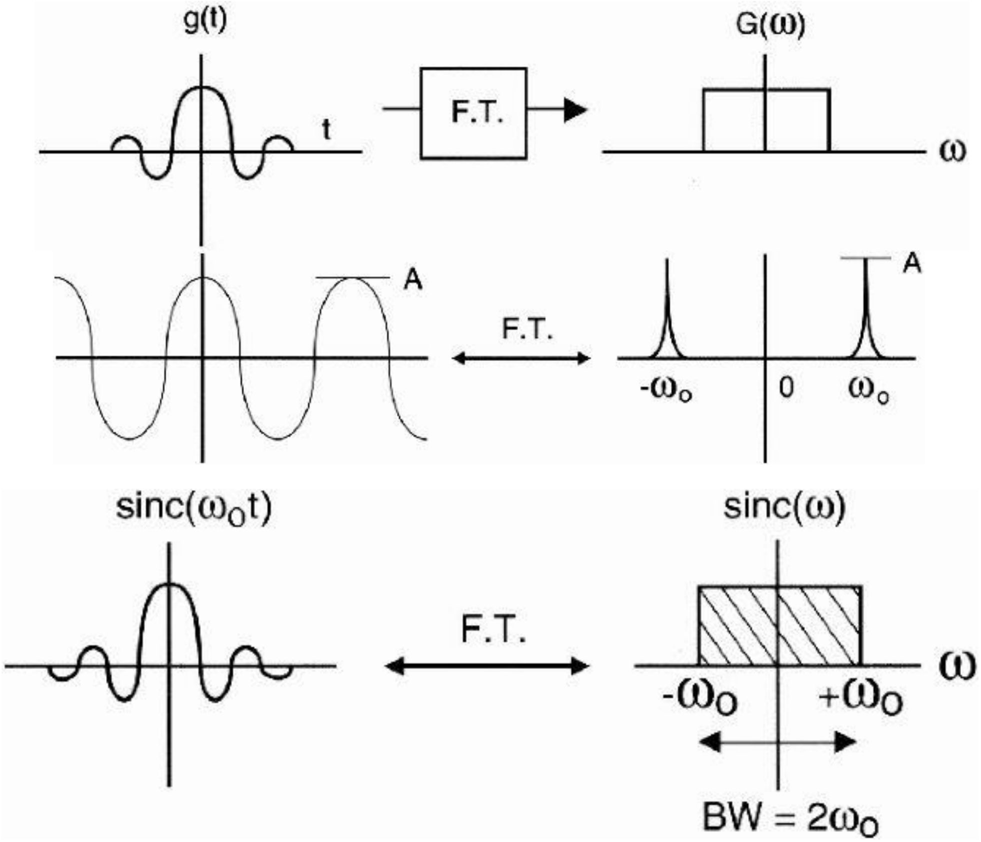
# Tissue Contrast with Spin Echo

| 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*)

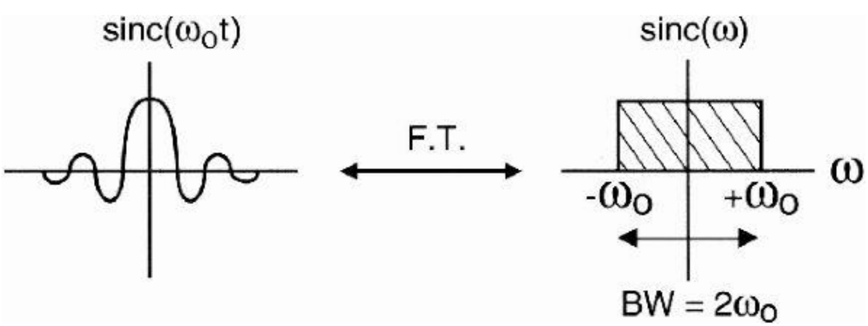
$$\mathcal{F}\{g\} = \iint\limits_{-\infty}^{\infty} g(x, y) \exp\left[-j2\pi(f_X x + f_Y y)\right] dx\, dy.$$

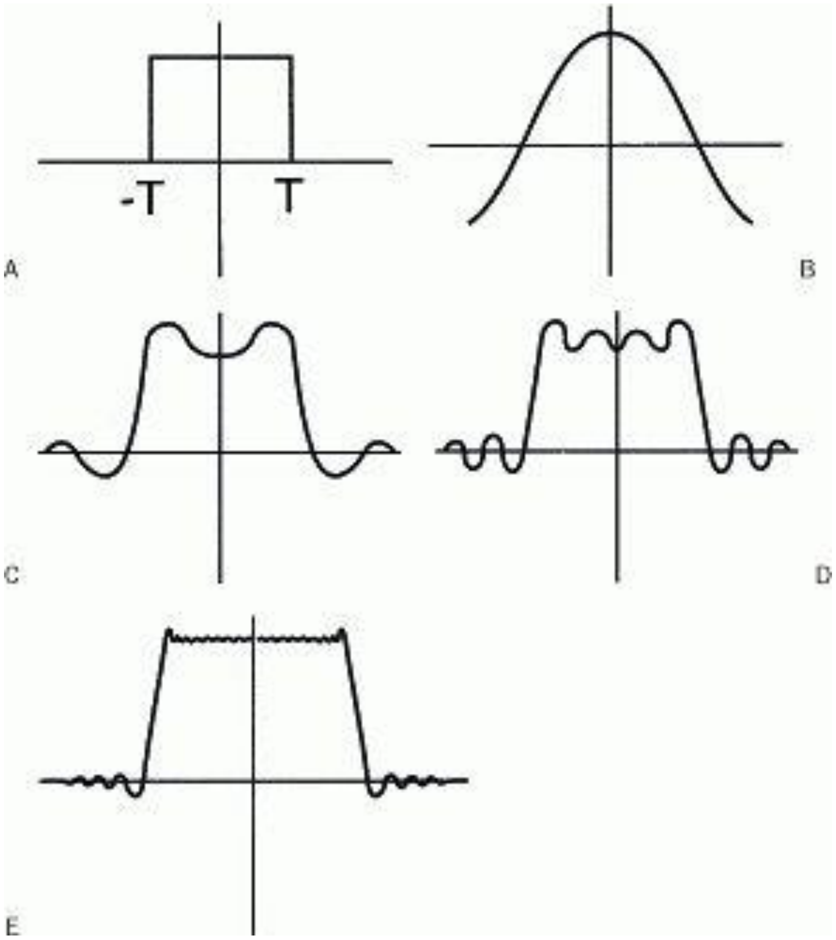
- Inverse transform (*Synthesis*)

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

# 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



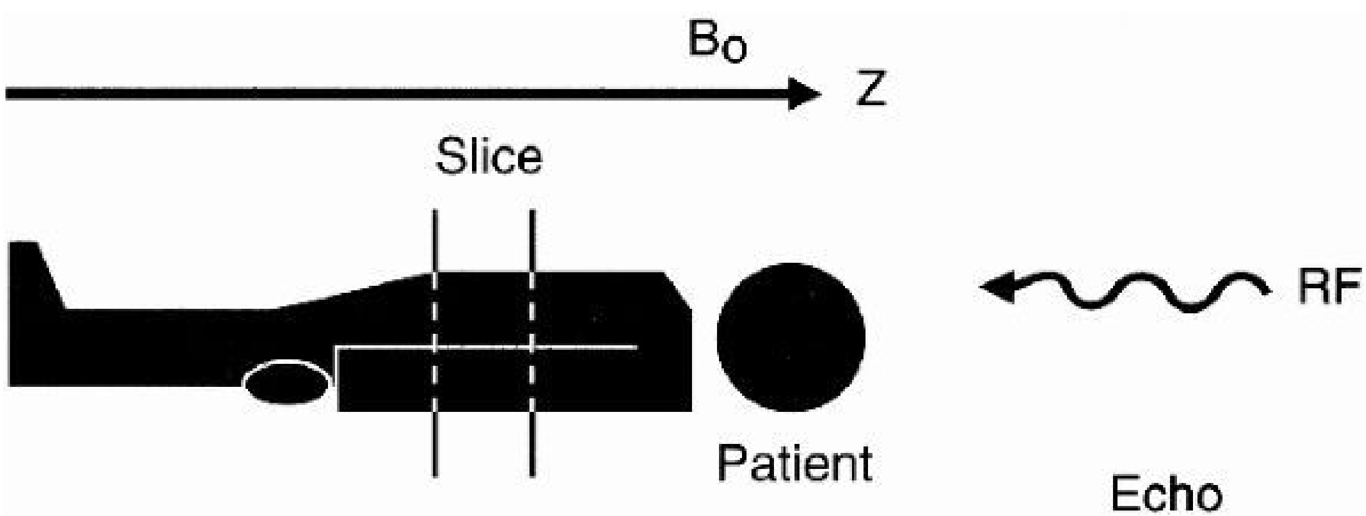



# Image Reconstruction

- 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

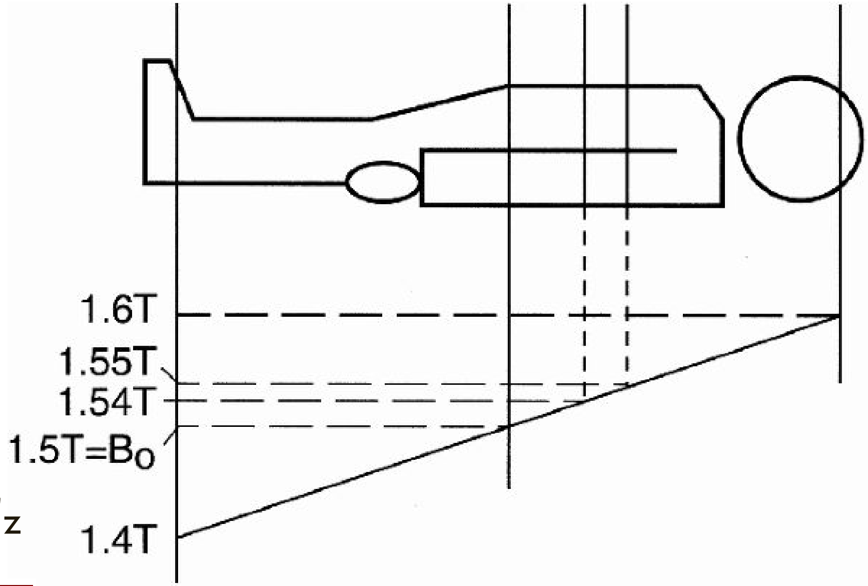
- Signal is obtained only from a particular slice from the body.
  - Can be in any direction



# Slice Selection

- Larmor equation:

$$\omega_o = \gamma B_0$$

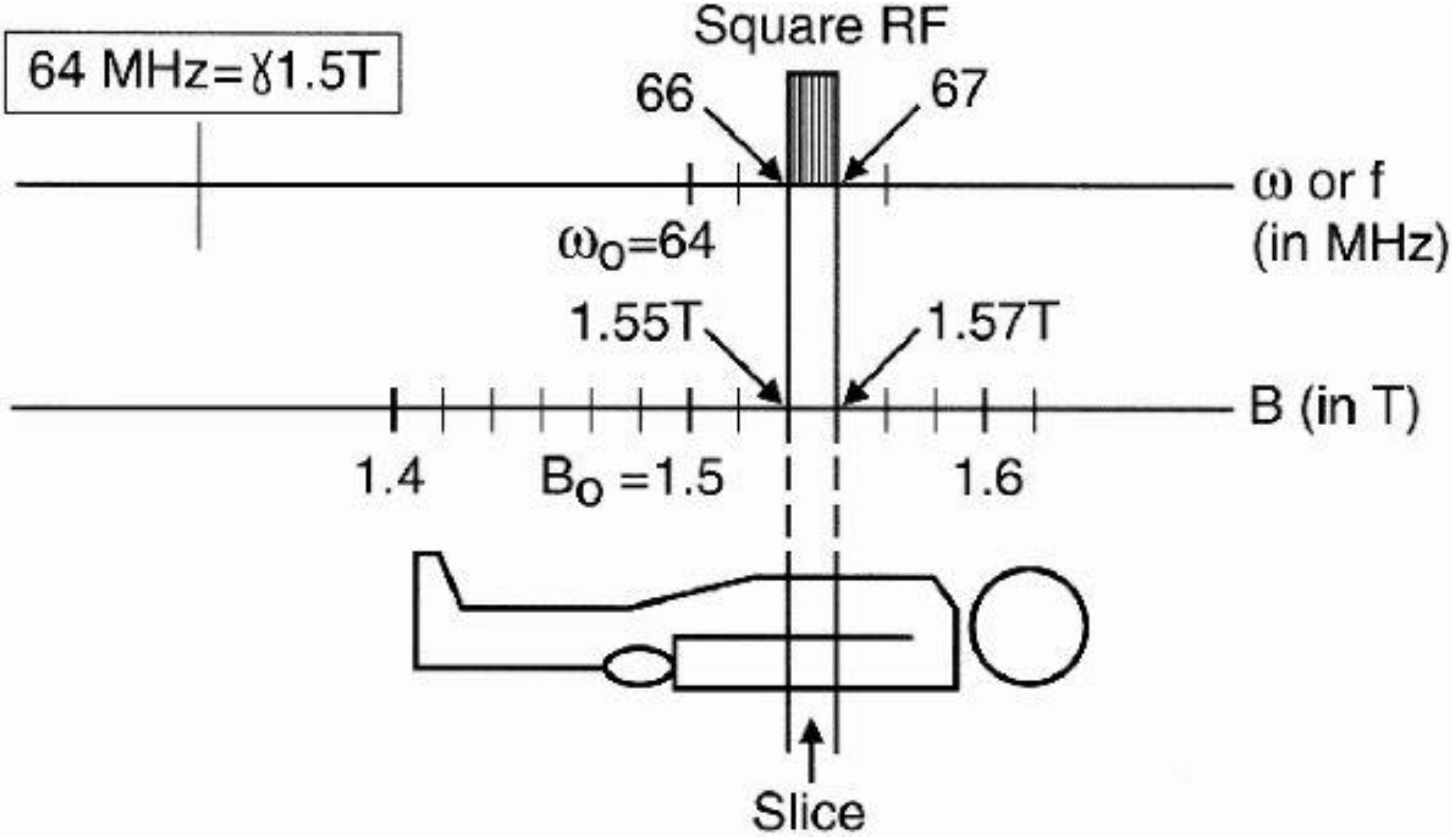



- Larmor equation with gradient $G_z$

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

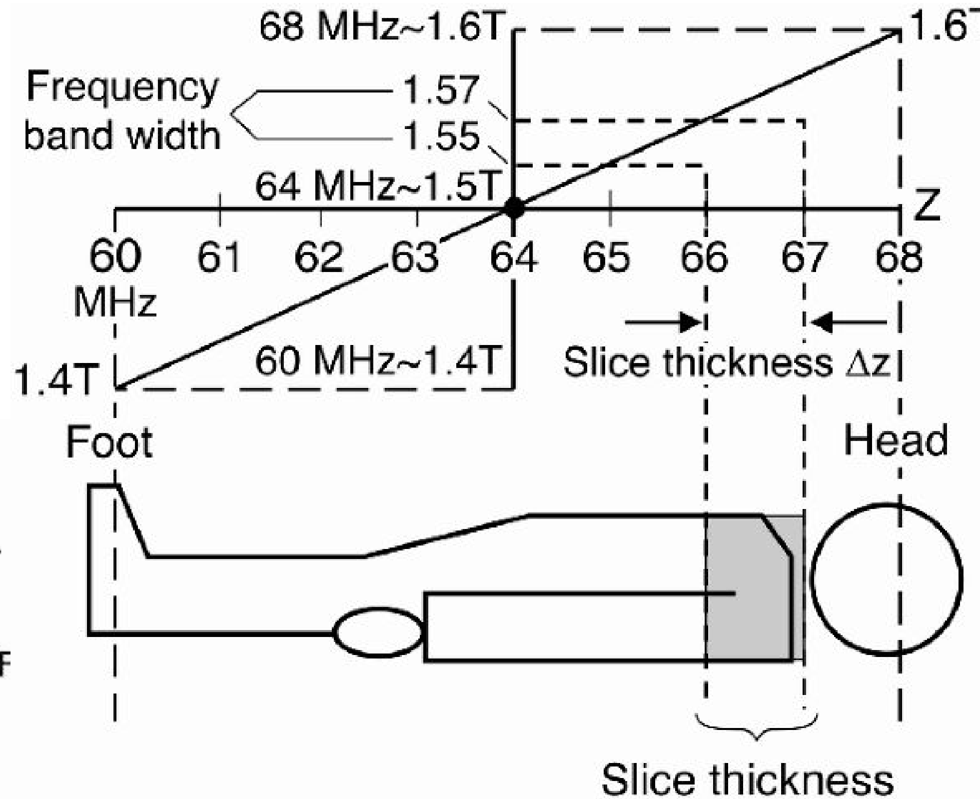
  - Larmor frequency depends on location
  - Send RF pulse with desired frequency range to excite a slice !

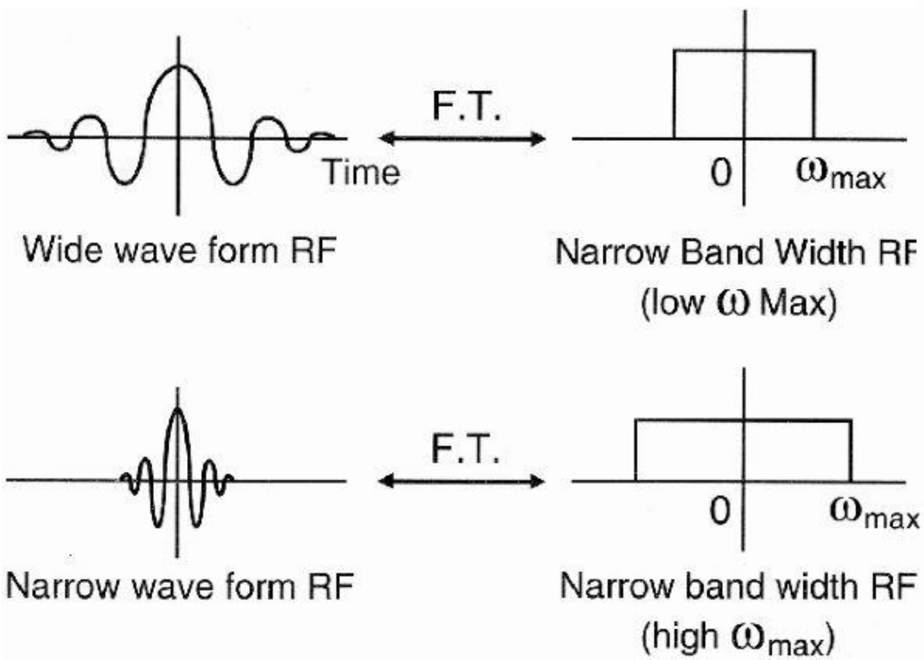
# Slice Selection



# Slice Selection

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

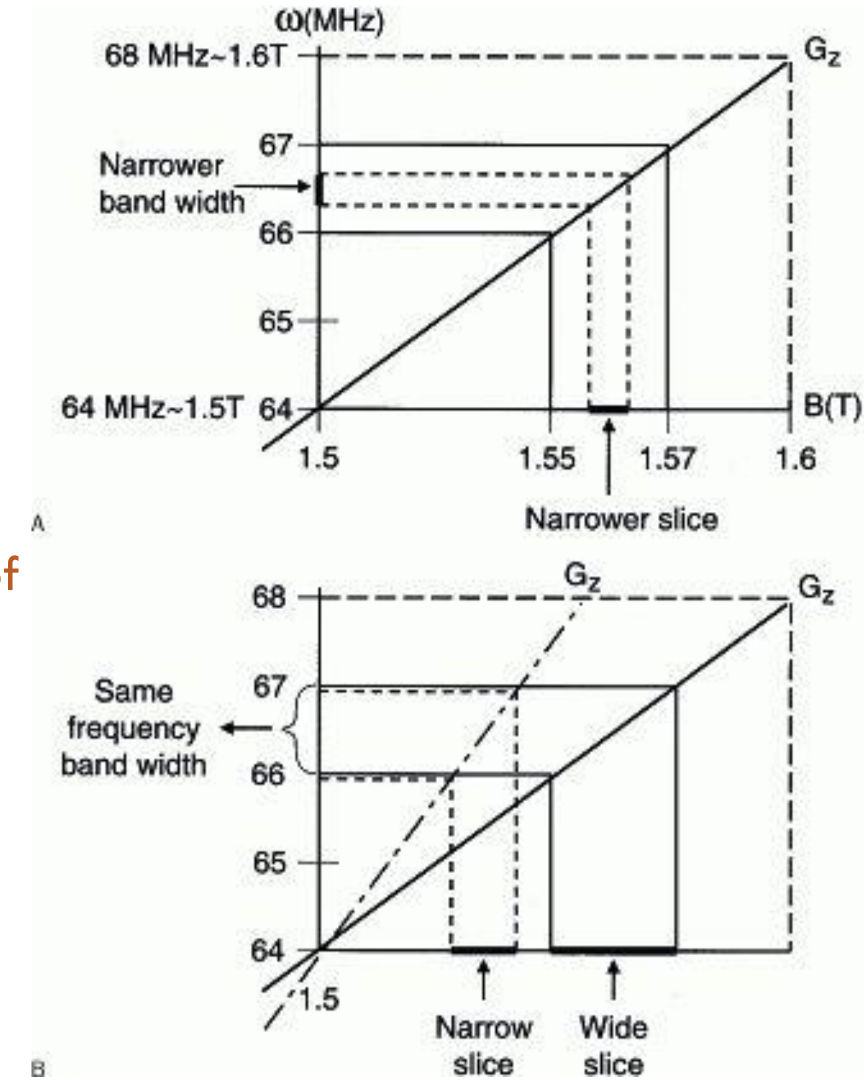


# Slice Selection: Changing Thickness

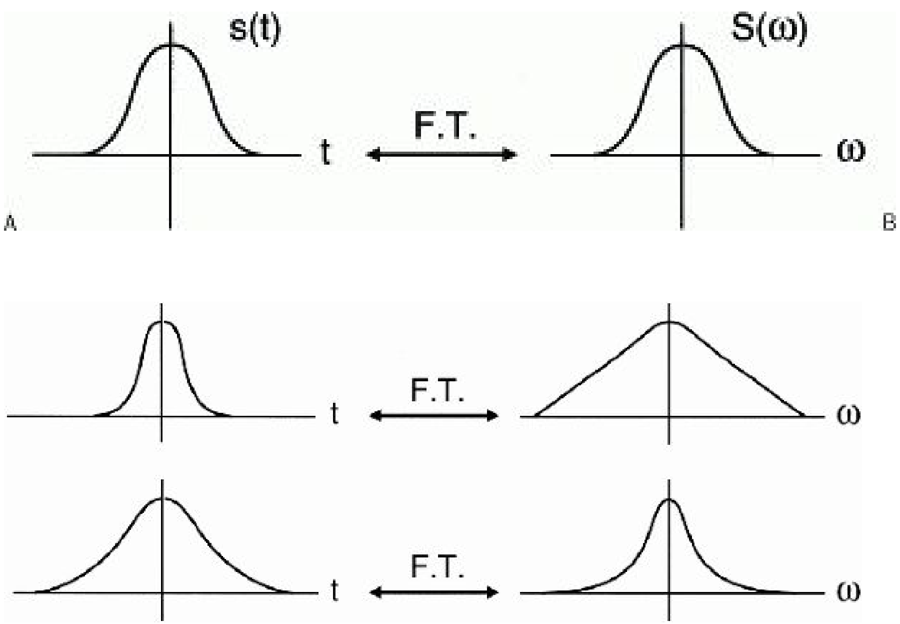
- Different RF pulse bandwidth
- Different slice selection gradient

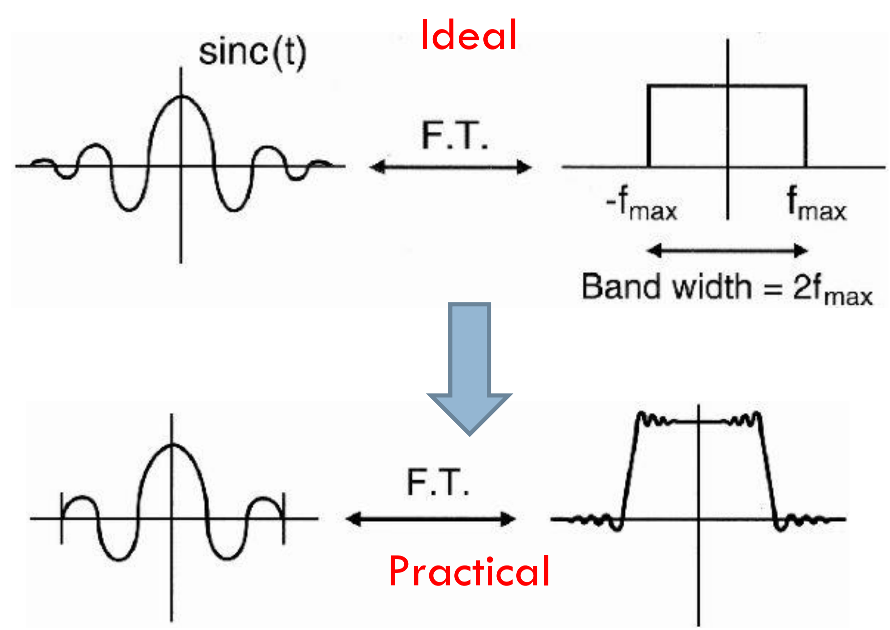
- To decrease the thickness is to use a narrower bandwidth.
  - Narrower frequency bandwidth will excite protons in a narrower band of magnetic field strengths
- Second way to decrease slice thickness is to increase the slope of the magnetic field gradient



# Slice Selection: RF Pulses

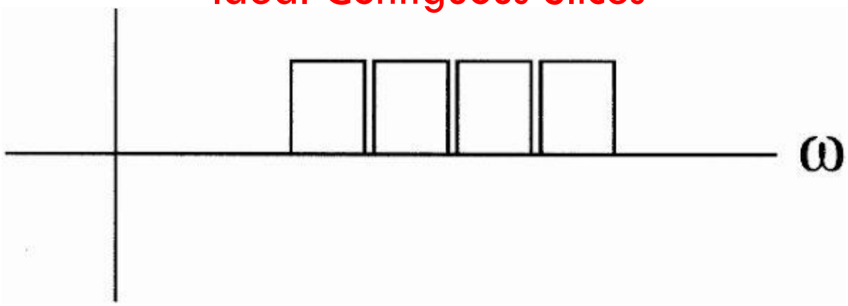
- There are two types of RF pulses:
  - Nonselective
  - Selective
- Slice profile = Fourier transform of pulse shape

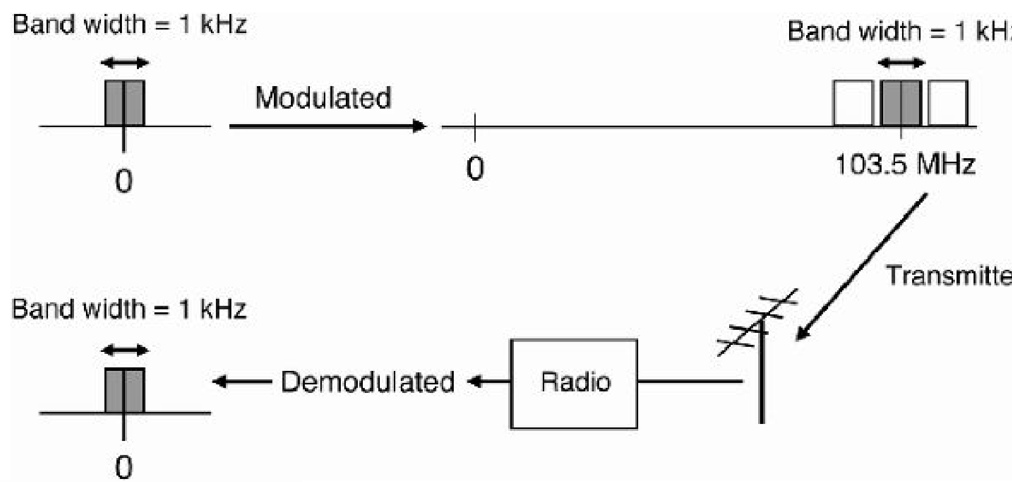


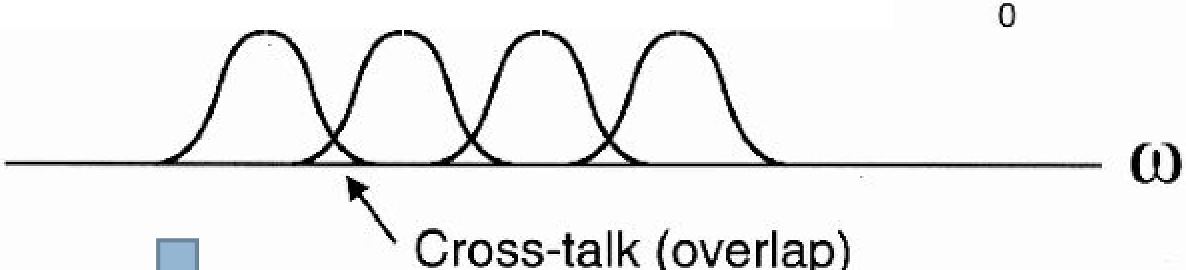
# Slice Selection: Multi-Slice Scan

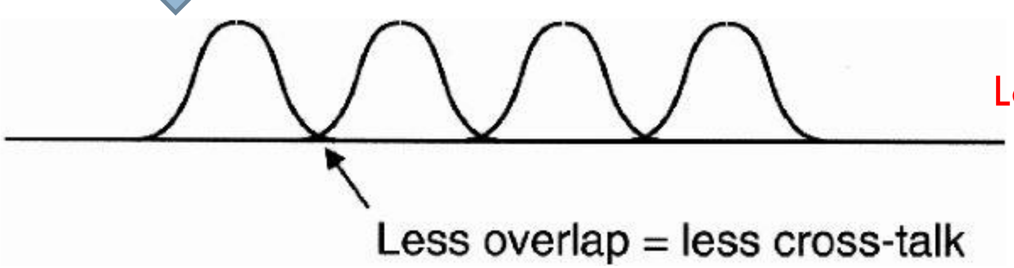
Ideal Contiguous Slices




Practical Slices




Leave more spacing between slices

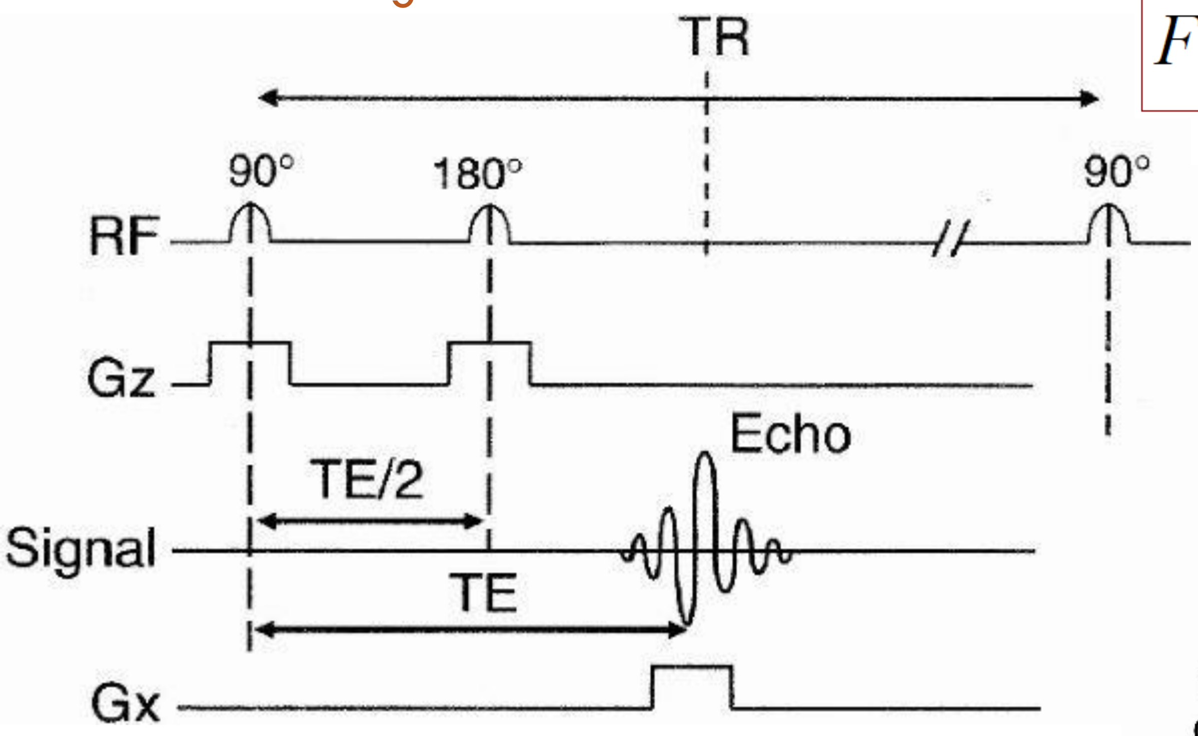
# In-Plane Spatial Encoding: Fourier Imaging

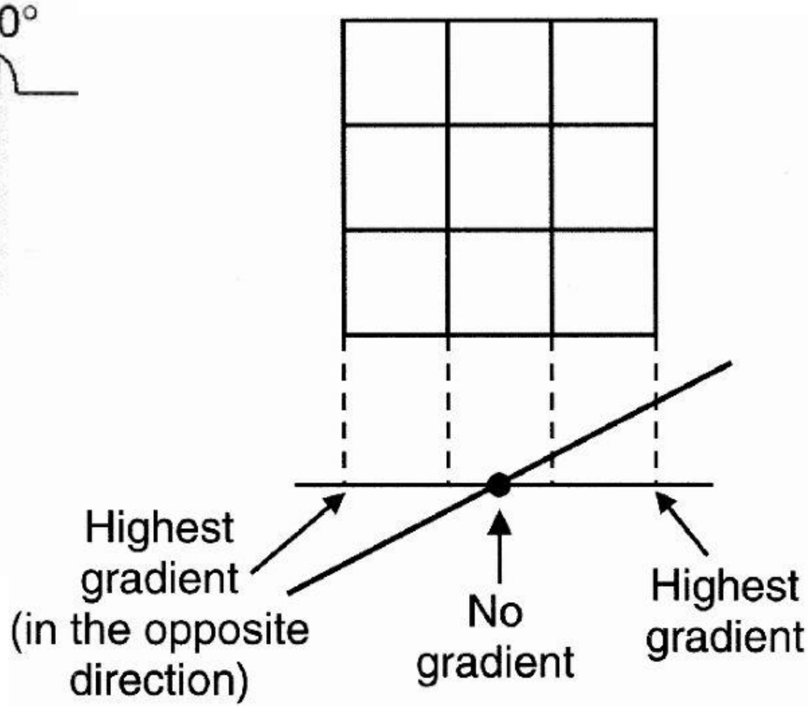
- Basic idea: encode location by frequency
  - Magnetic field gradient is used during reception
  - Larmor frequency depends on present magnetic field
  - Returned frequency from an area depends on its location
  - Easily decoded by Fourier transformation
- Applied by 2 different methods
  - Frequency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

| | | |
|---|---|---|
| 0 | 1 | 1 |
| 1 | 2 | 0 |
| -2 | 0 | 1 |

True Image

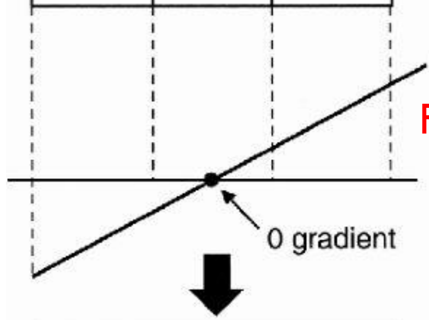
| | | |
|---|---|---|
| 0 | $\cos\omega t$ | $\cos\omega t$ |
| $\cos\omega t$ | $2\cos\omega t$ | 0 |
| $-2\cos\omega t$ | 0 | $\cos\omega t$ |

Image sum

→ $4\cos\omega t$

No Frequency Encoding

| | | |
|---|---|---|
| 0 | $\cos\omega t$ | $\cos\omega t$ |
| $\cos\omega t$ | $2\cos\omega t$ | 0 |
| $-2\cos\omega t$ | 0 | $\cos\omega t$ |




Frequency Encoding Applied

| | | |
|---|---|---|
| 0 | $\cos\omega t$ | $\cos\omega o^{+}t$ |
| $\cos\omega o^{-}t$ | $2\cos\omega t$ | 0 |
| $-2\cos\omega o^{-}t$ | 0 | $\cos\omega o^{+}t$ |

Column sum

| | | |
|---|---|---|
| $-\cos\omega o^{-}t$ | $3\cos\omega t$ | $2\cos\omega o^{+}t$ |

→ $(-\cos\omega o^{-}t) + (3\cos\omega t) + (2\cos\omega o^{+}t)$

# Phase Encoding

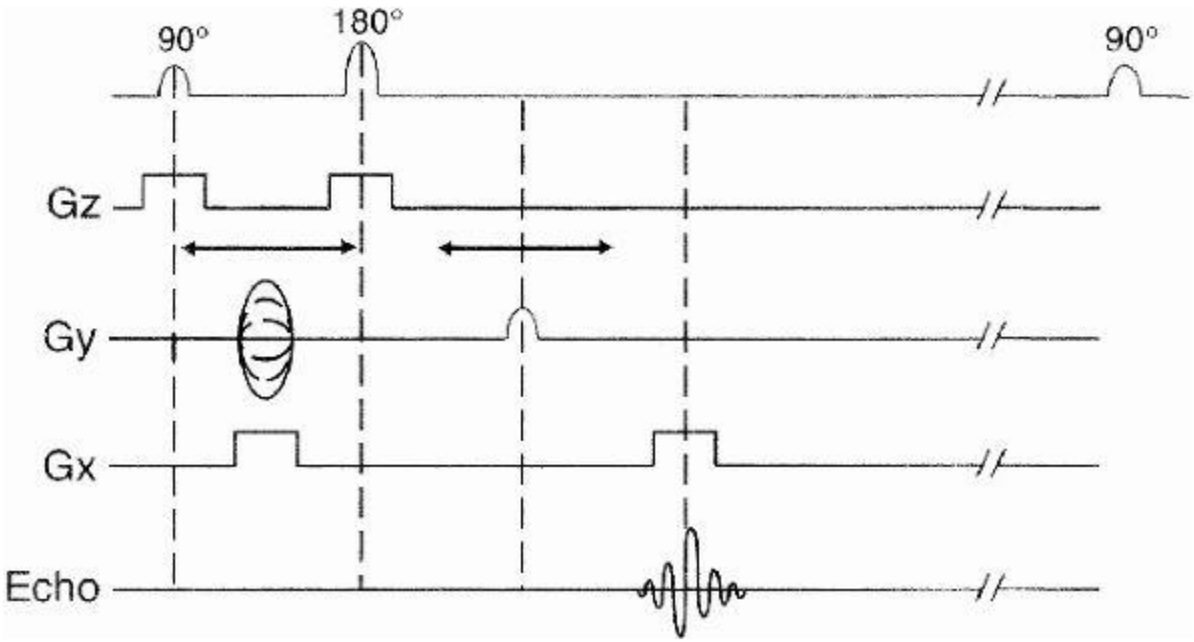
- Can we apply frequency encoding in 2 directions simultaneously?
  - Answer is NO
- 2D Fourier transform

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

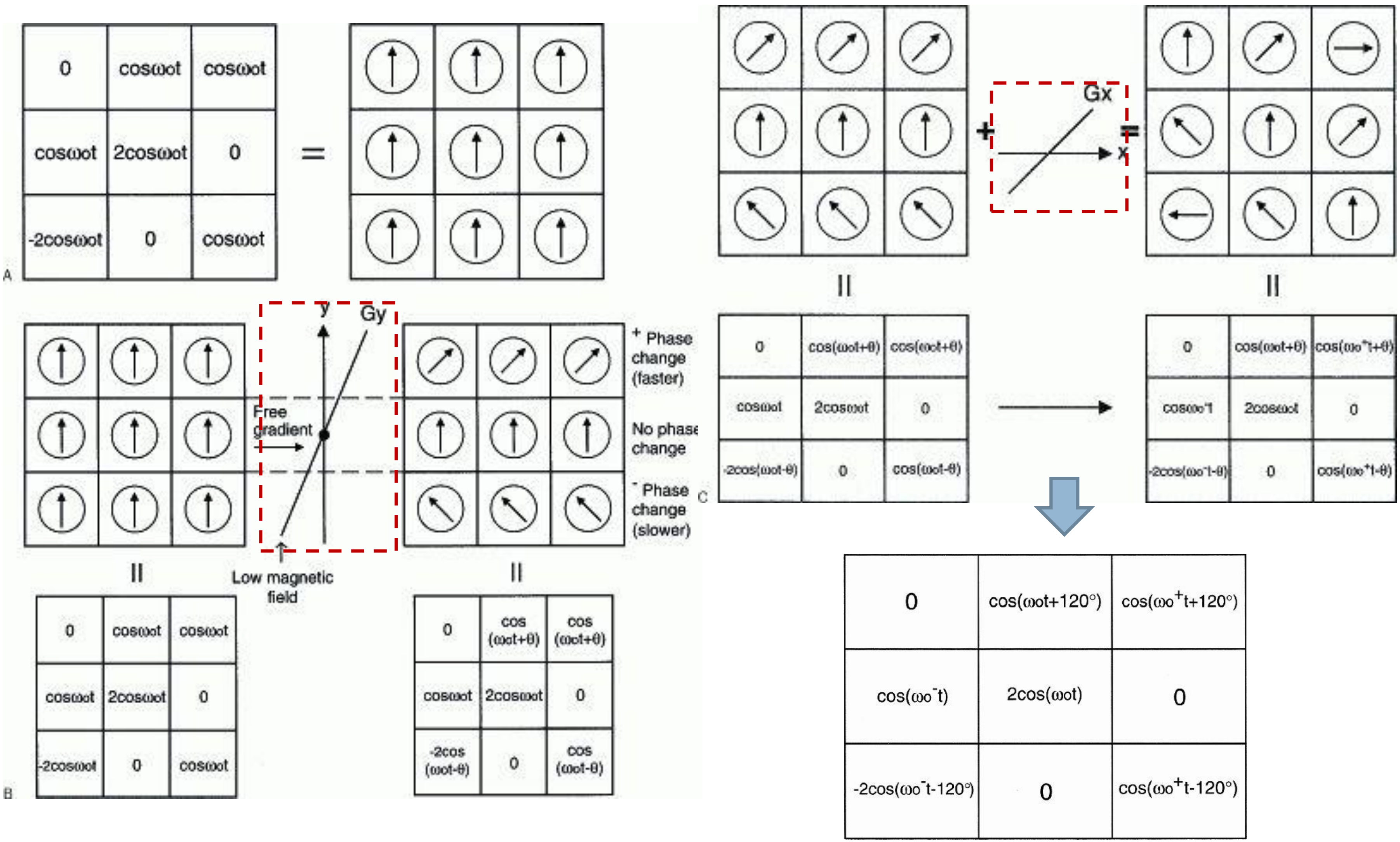
$$F(k_x, k_y) = \underbrace{\int e^{-j2\pi k_y y}}_{\text{Phase Encoding}} \underbrace{\left\{ \int f(x, y) \cdot e^{-j2\pi k_x x} dx \right\}}_{\text{Frequency Encoding}} dy$$

# Phase Encoding

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



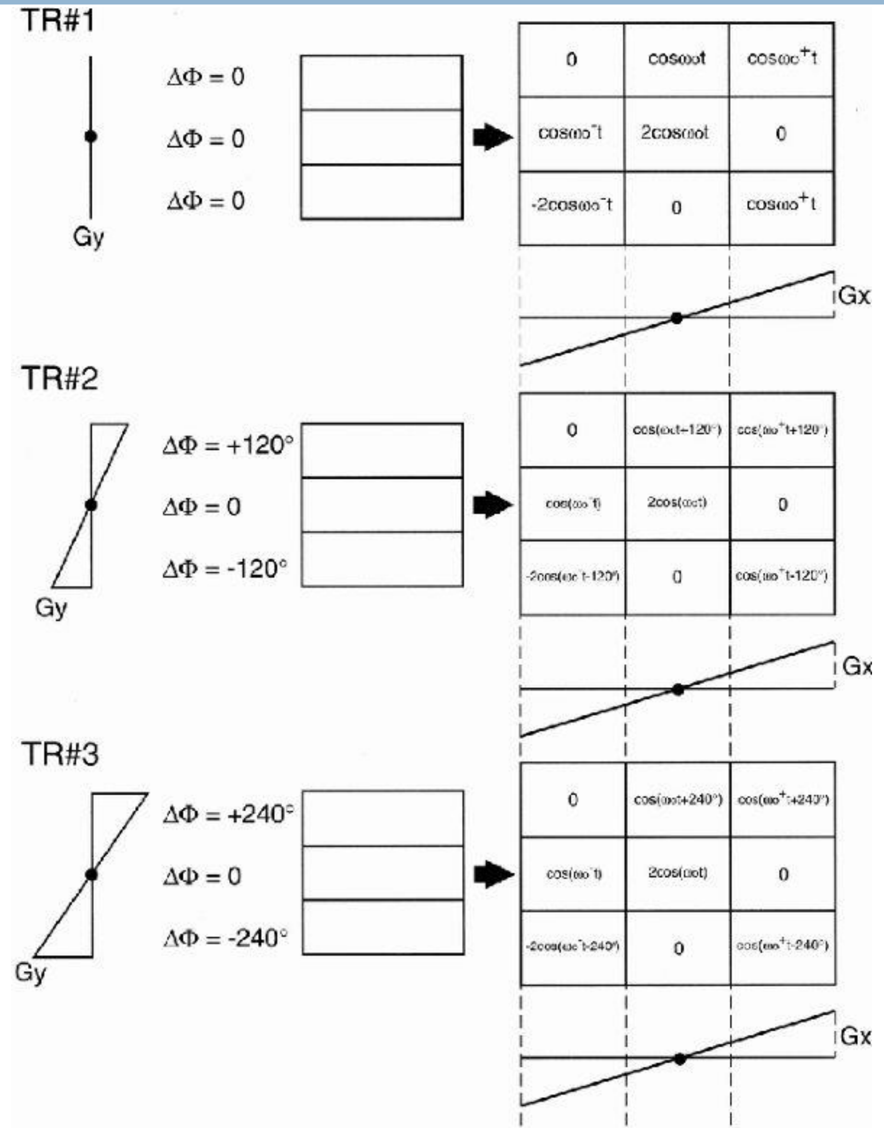
# Phase Encoding



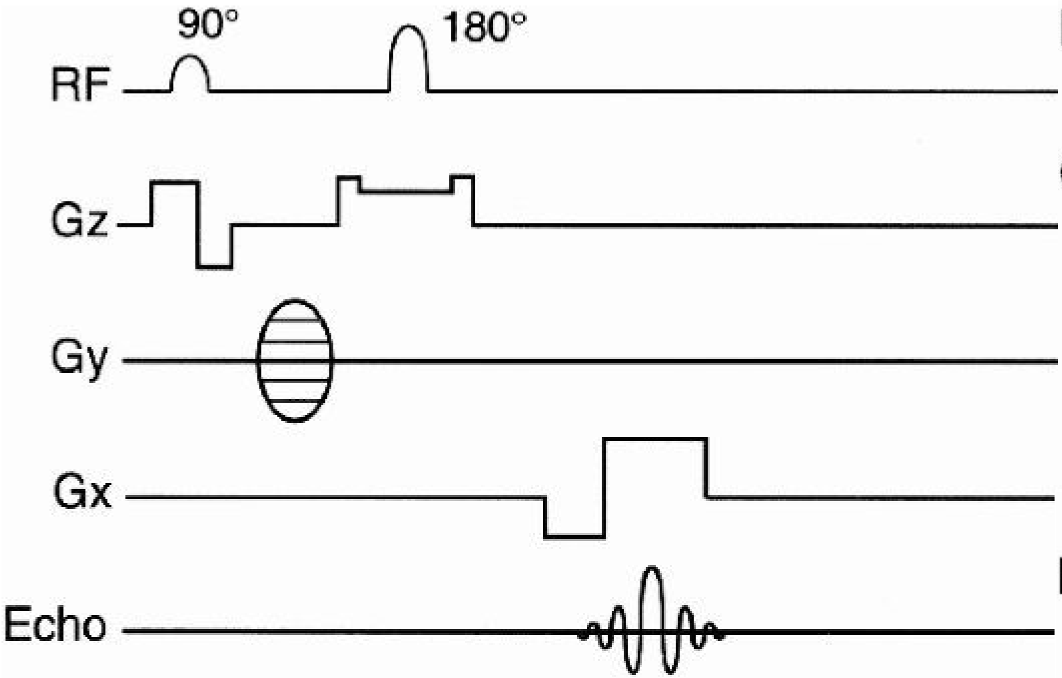
# Phase Encoding

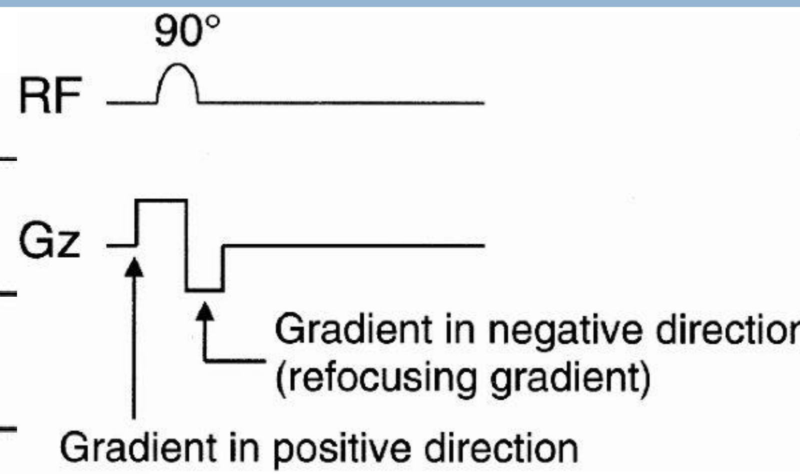
- Each phase encoding requires 1 RF pulse
  - Acquisition time = #phase encoding steps x TR

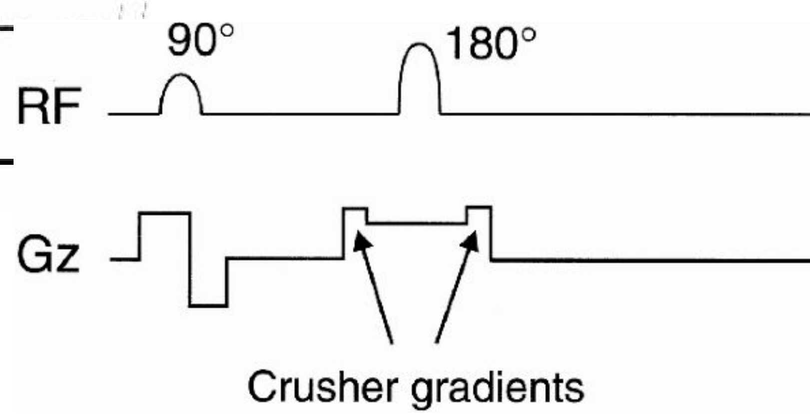
*The protons in each pixel have a distinct frequency and a distinct phase, which are unique and encode for the x and y coordinates for that pixel.*



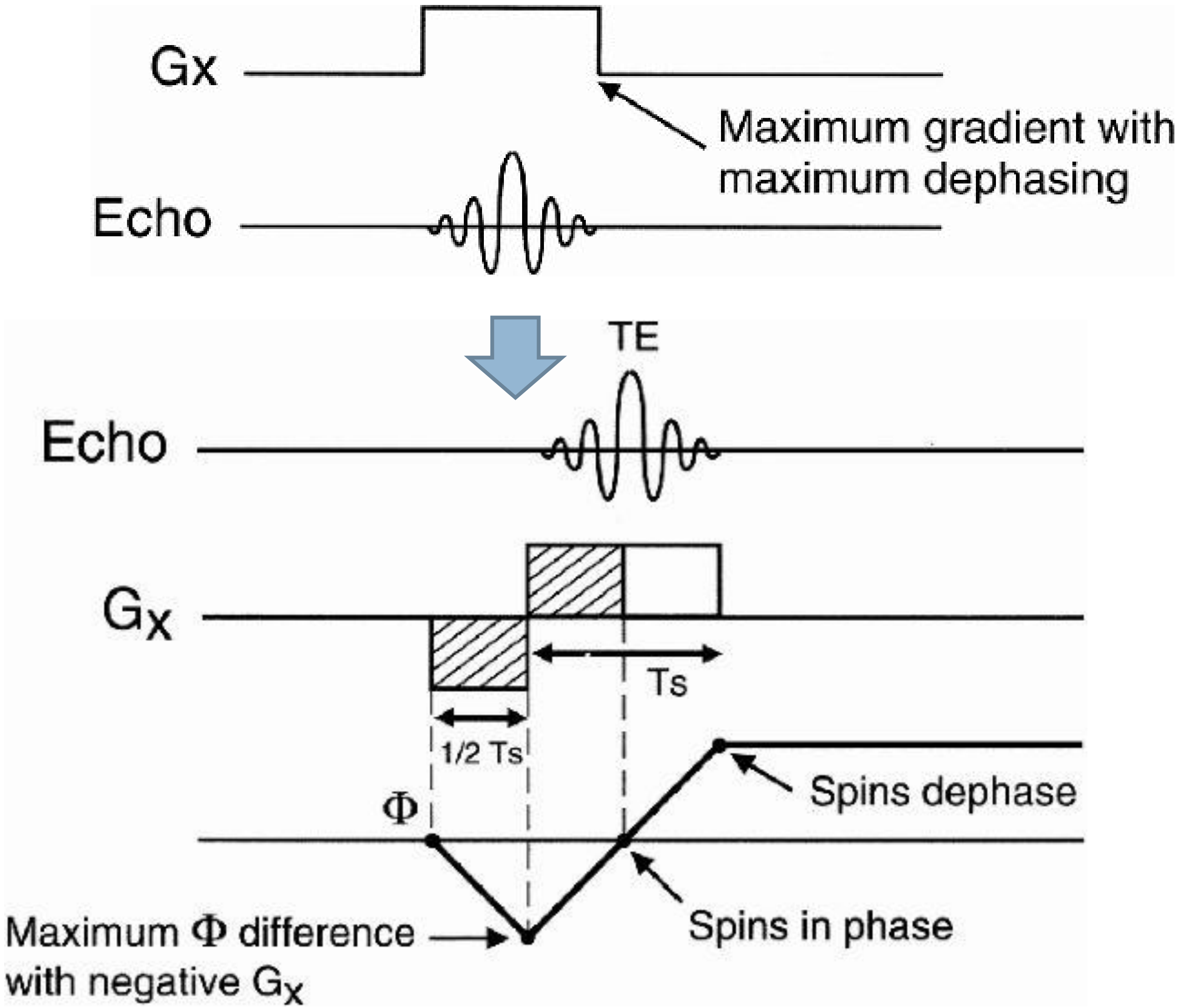
# Pulse Sequence Diagram



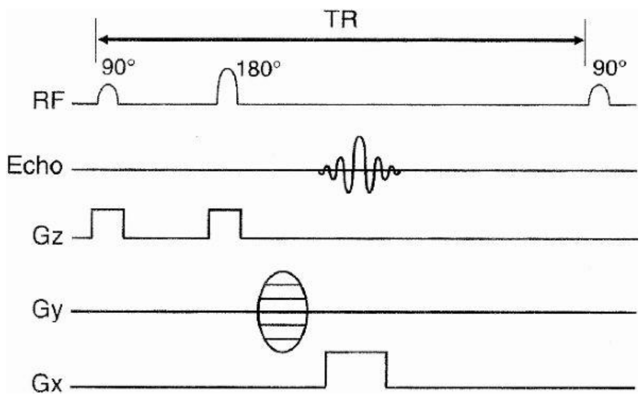
# Pulse Sequence Diagram

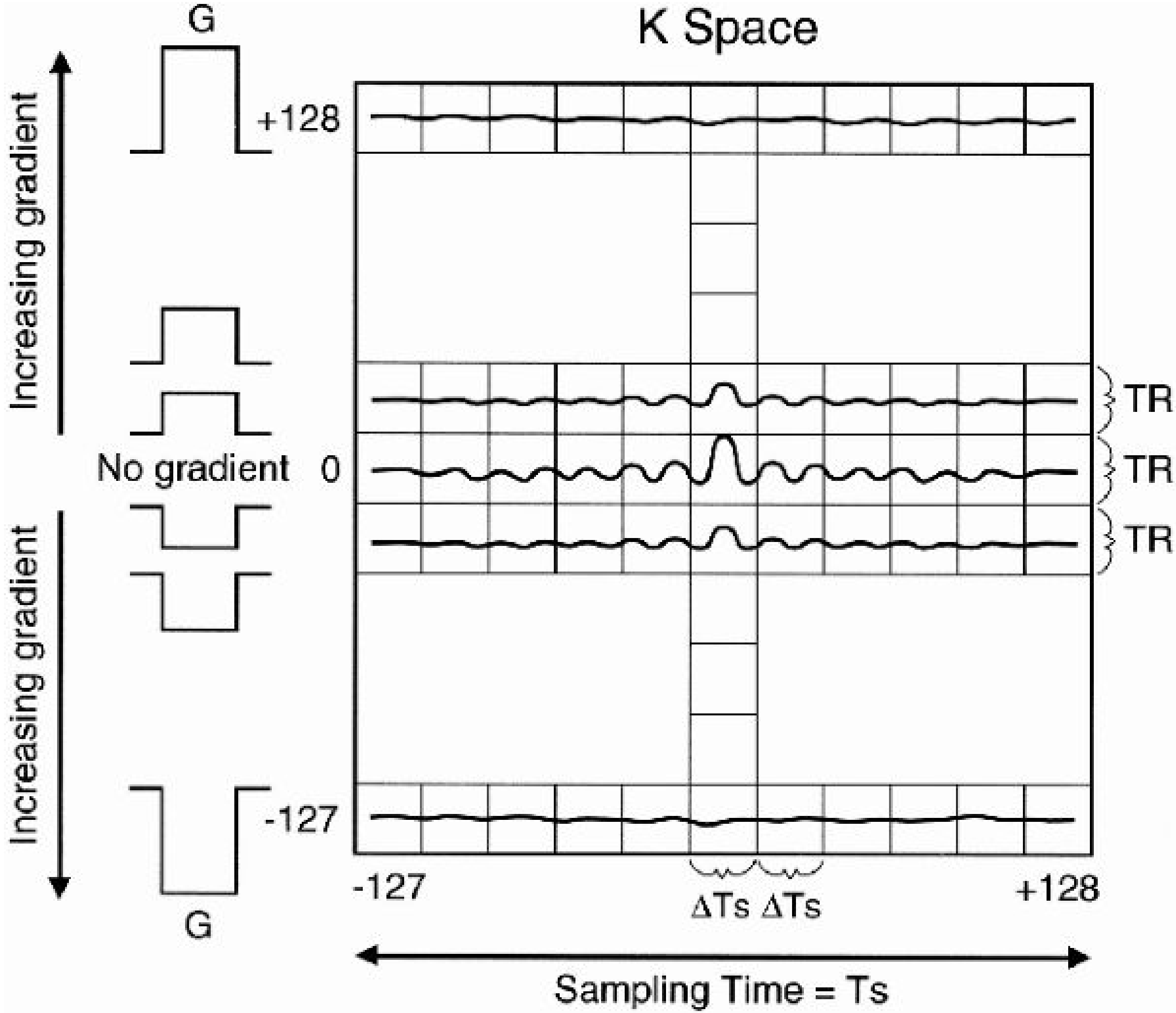


# K-Space and Image Space

- K-space= Fourier domain

Sample Spin-Echo Pulse Sequence

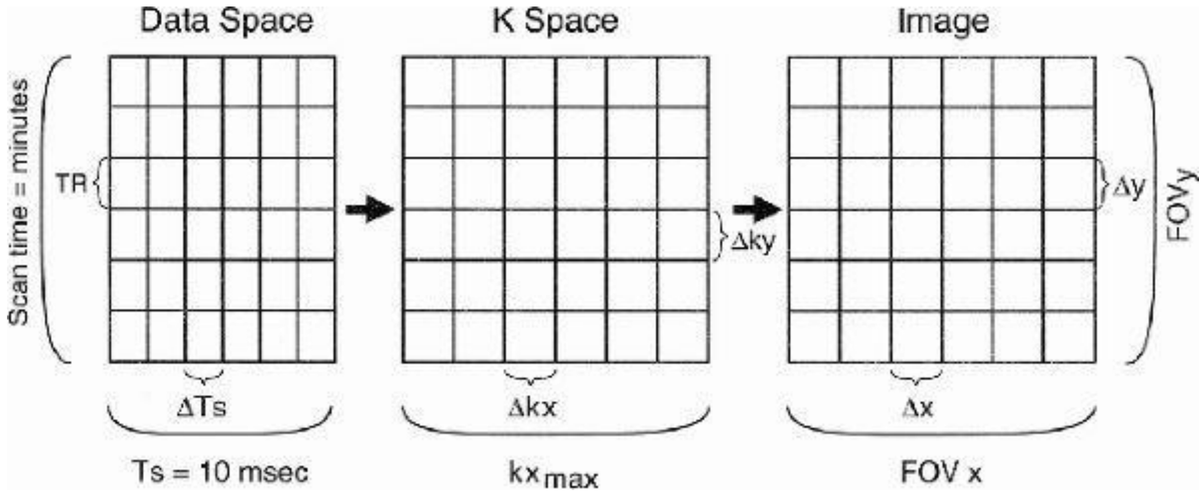


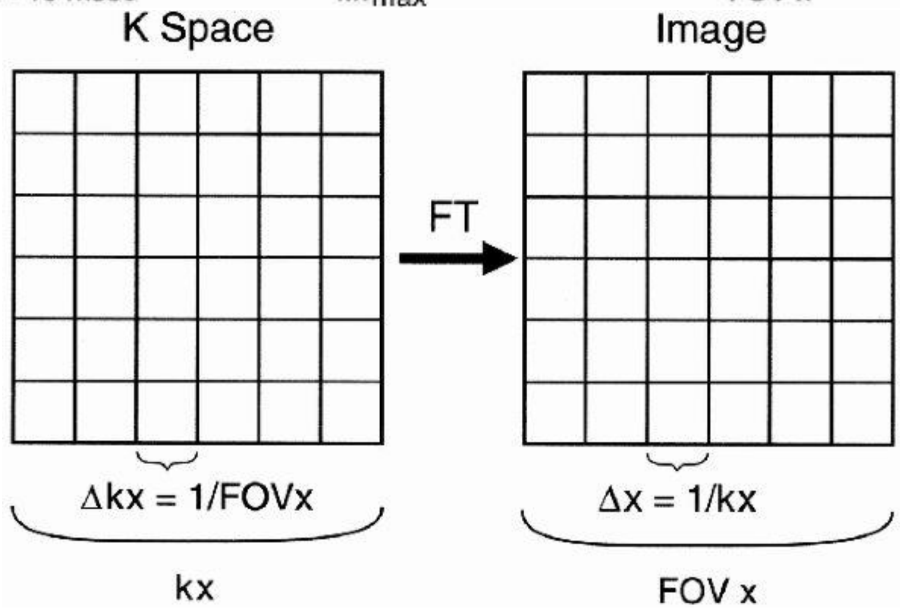



# K-Space and Image Space

- 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*
- $\Delta z$ = 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:* $\Delta x$ : spacing in x direction; $\Delta y$ : spacing in y direction
- NEX, Bandwidth
- All of the above (FOV, $N_x$, $N_y$, 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)

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

- 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:

$$\text{SNR (3D)} = (FOV_x/N_x)(FOV_y)(FOV_z)\sqrt{\frac{NEX}{(N_y)(N_z)(BW)}}$$

# Parameter Optimization

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

$$\text{Scan time} = TR \cdot Ny \cdot Nz \cdot NEX$$

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

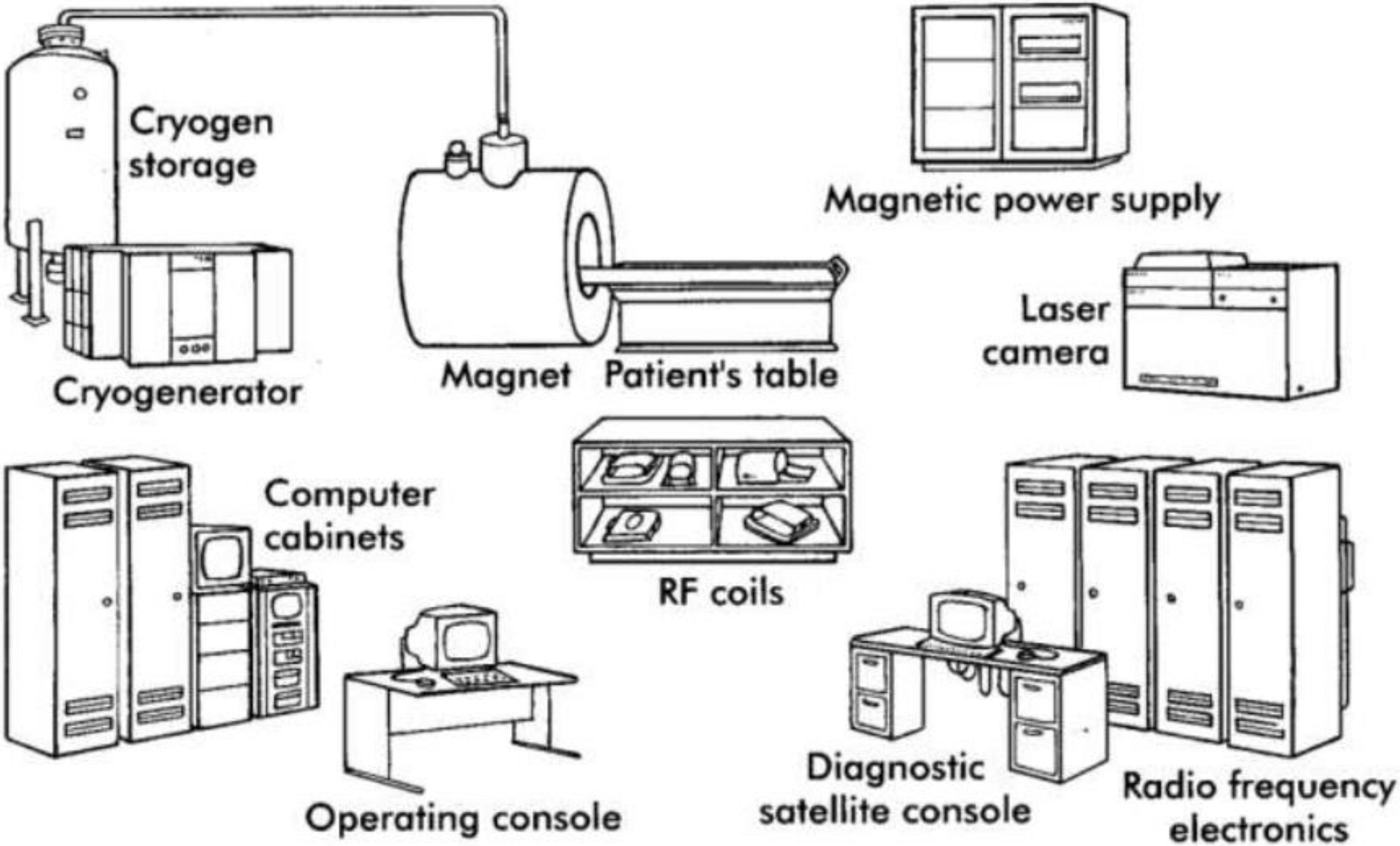
# Parameter Optimization: Examples

- If we keep FOV constant and increase $N_y$, we will decrease SNR. ↑ $N_y$, FOV constant → ↓ SNR
- If we increase $N_y$ and increase FOV, thus keeping pixel size constant, then we will increase the SNR.
  - ↑ FOV, pixels fixed → ↑ SNR, ↑ 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 $\Delta Ts$ (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 $\Delta Ts$ (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).

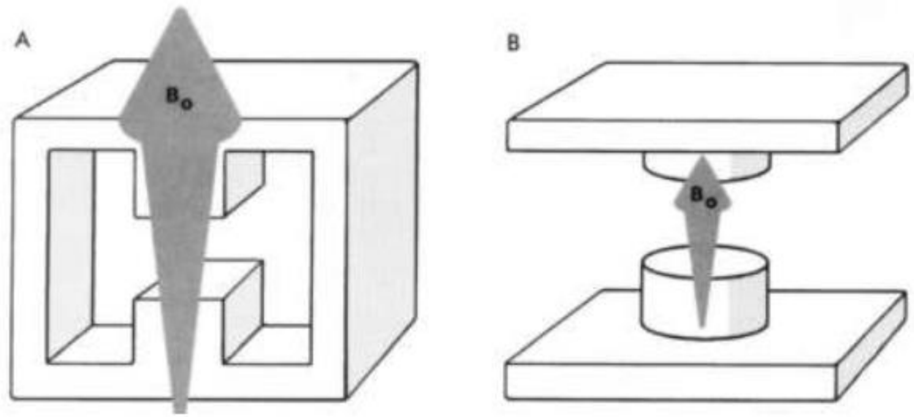
# Block Diagram of MRI System



# Primary Magnetic Field (B0)

- Permanent magnet
- Resistive magnet
- Superconductive magnet

# Permanent Magnet



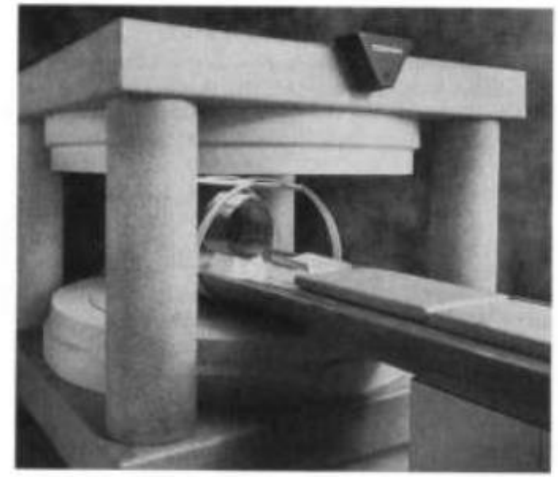
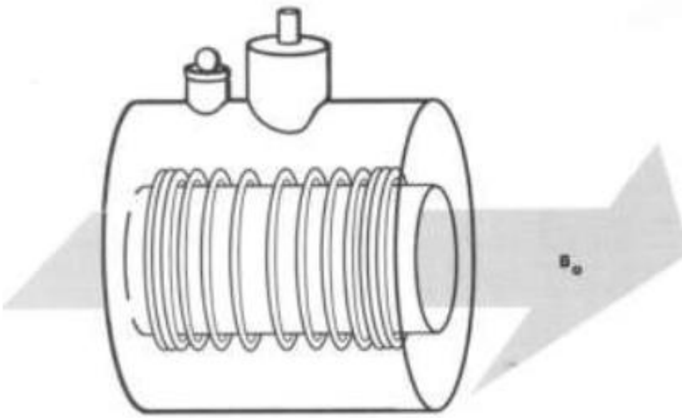



**Table 11-1** Characteristics of a permanent magnet magnetic resonance imager

| 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



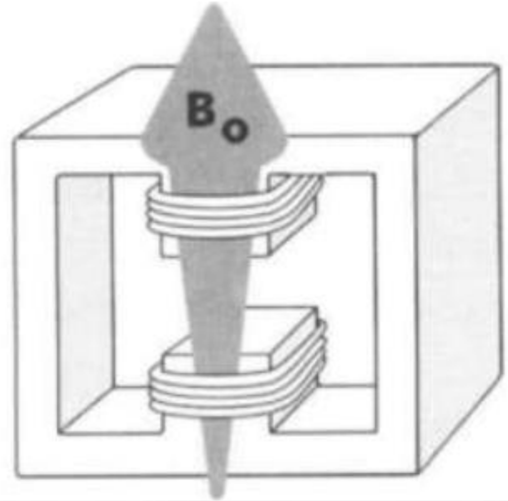
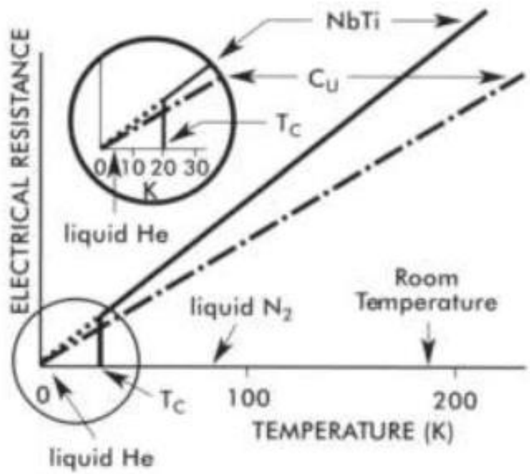



**Table 11-2** **Characteristics of a resistive electromagnet MR imager**

| 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



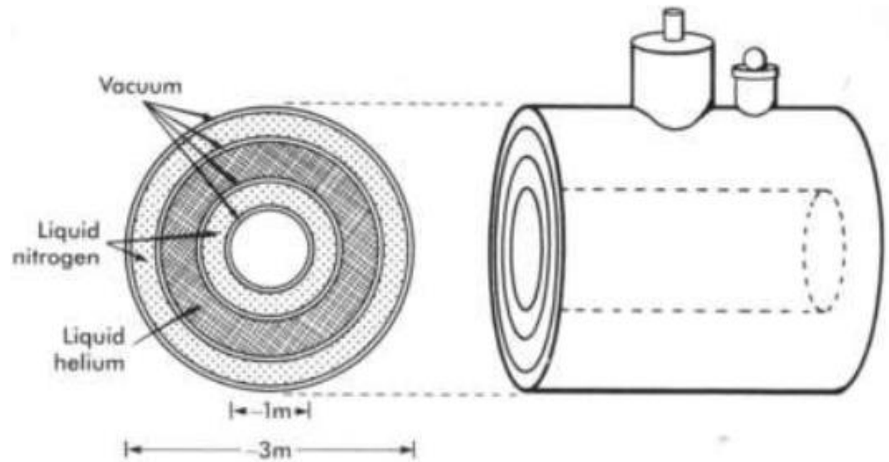
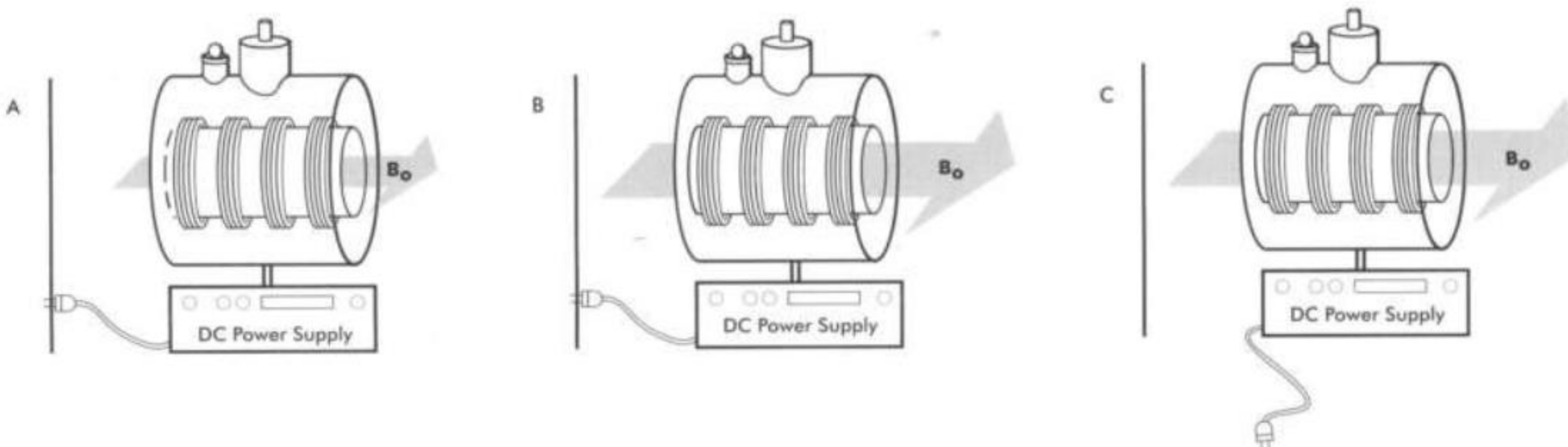




**Table 11-3** Characteristics of a superconducting electromagnet magnetic resonance imager

| 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

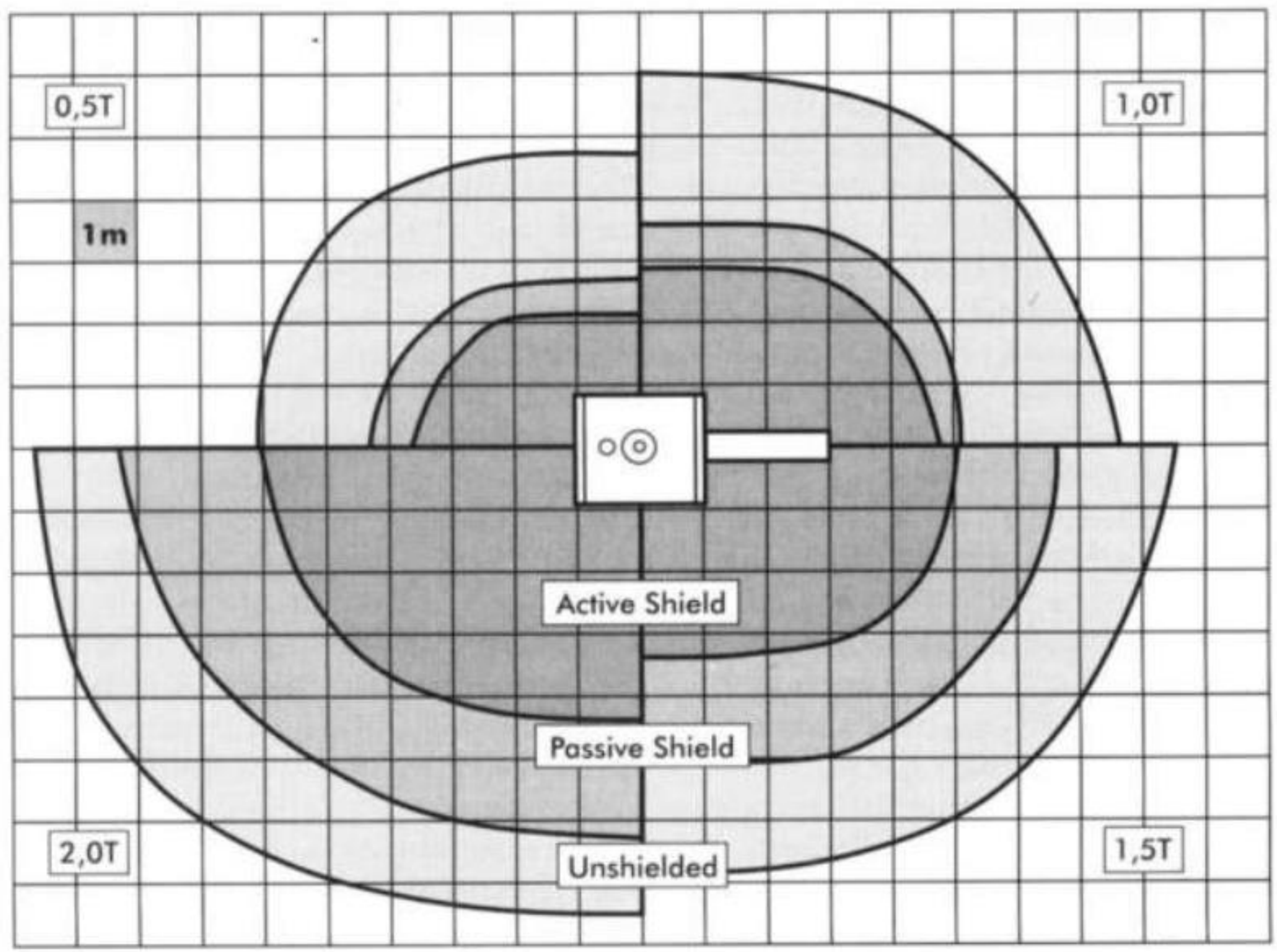
- Magnetic field ramp-up




- 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



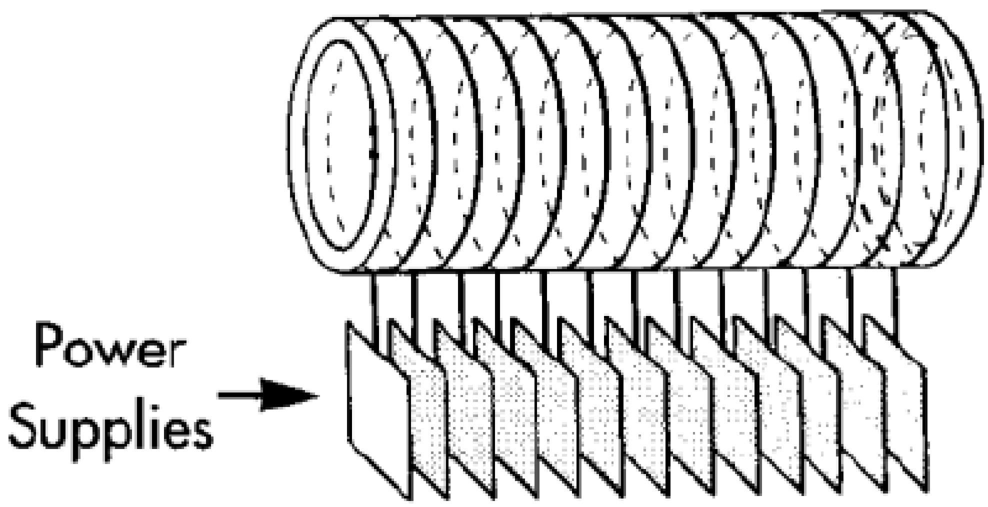

Distance to Safe 5G (0.5 mT) Line for different fields

# Secondary Magnets: Coils

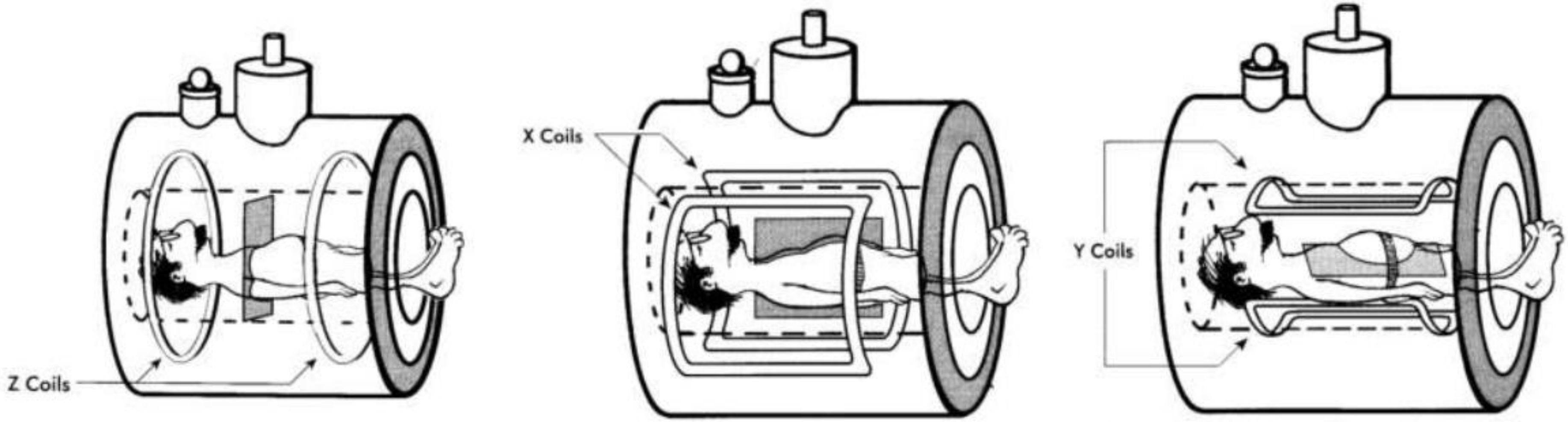
- Shim coils
  - Improves B0 field uniformity to within a few **ppm scale**
- Gradient coils
  - Apply gradients in x, y, and z directions for slice selection, frequency and phase encoding.
- RF coils
  - Send RF pulses and receive signal from patient

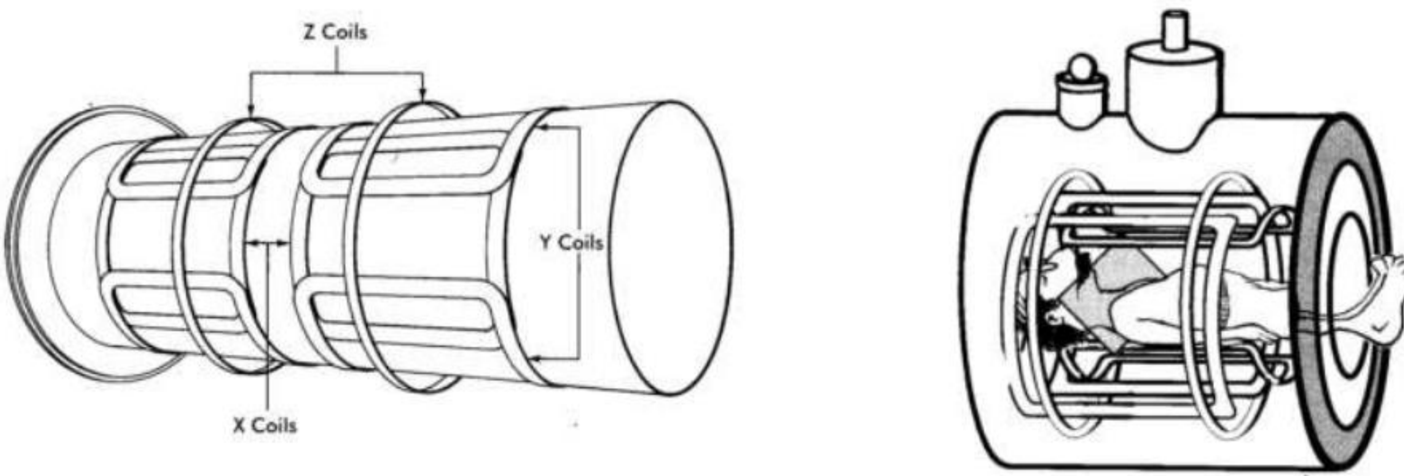
# Shim Coils

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



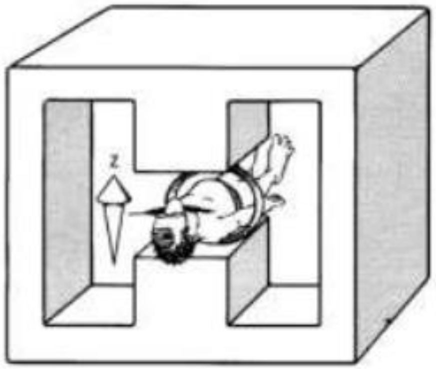
# Gradient Coils

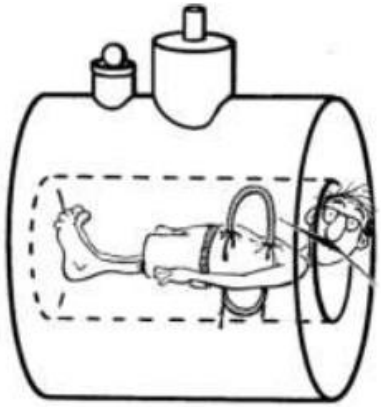


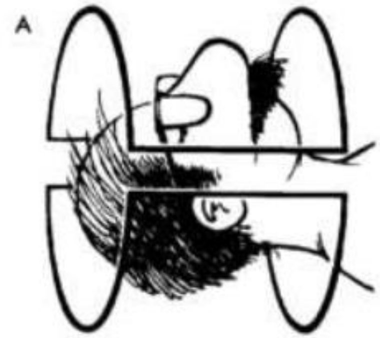



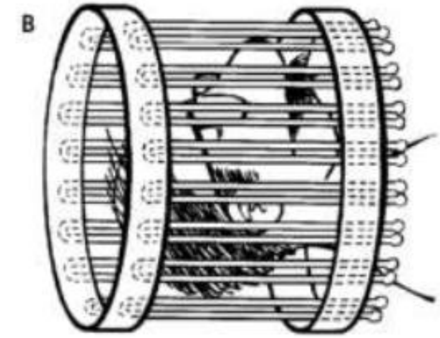

- Units: mT/m
- Slew rate
- Limited for safety

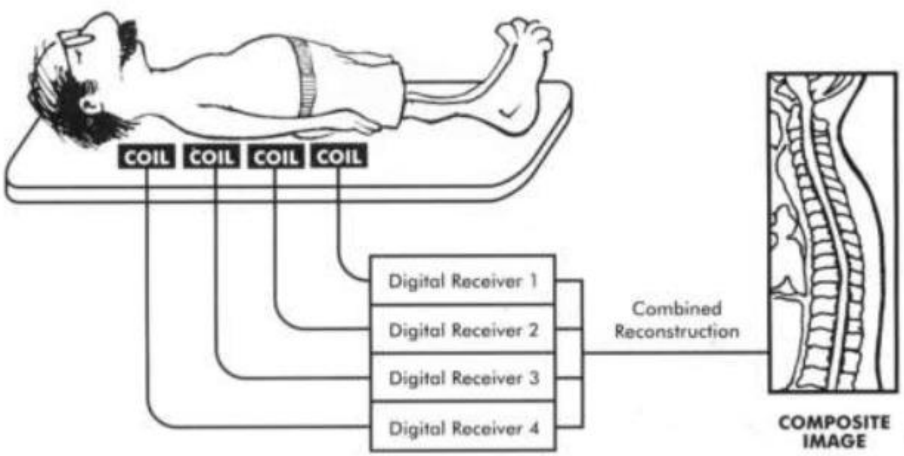
# RF Coils

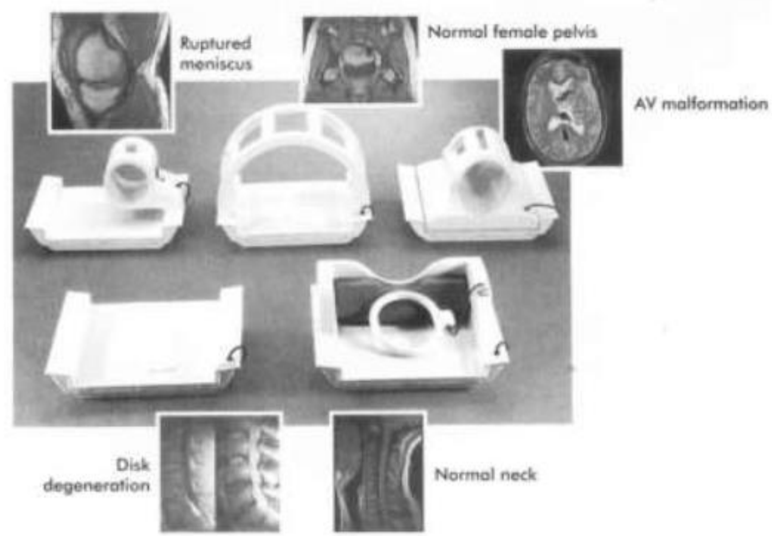





A


B

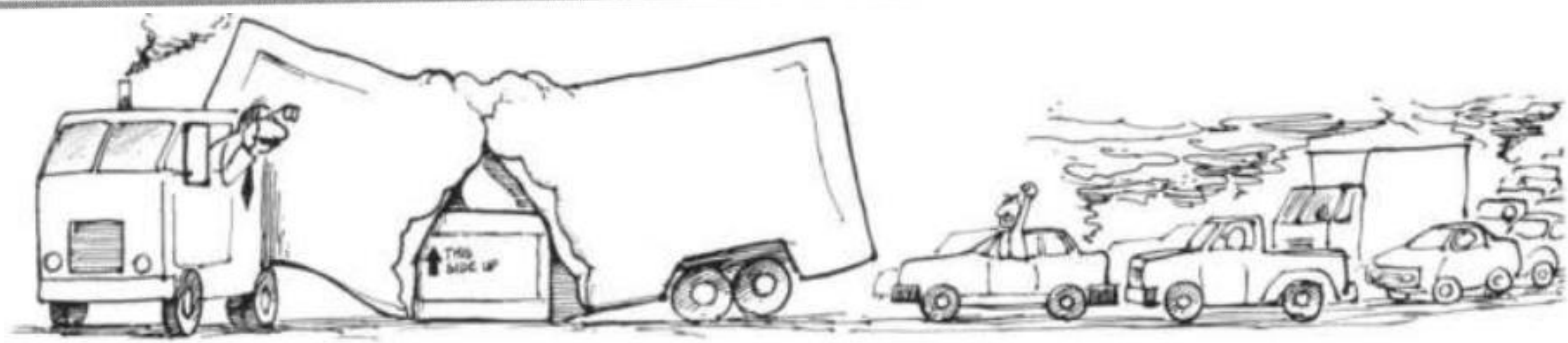

COIL
COIL
COIL
COIL
Digital Receiver 1
Digital Receiver 2
Digital Receiver 3
Digital Receiver 4
Combined Reconstruction
COMPOSITE IMAGE


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) | $1.5 \times 2.0$ | $2.1 \times 2.3$ | $2.3 \times 3.0$ |
| Weight (kg × 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

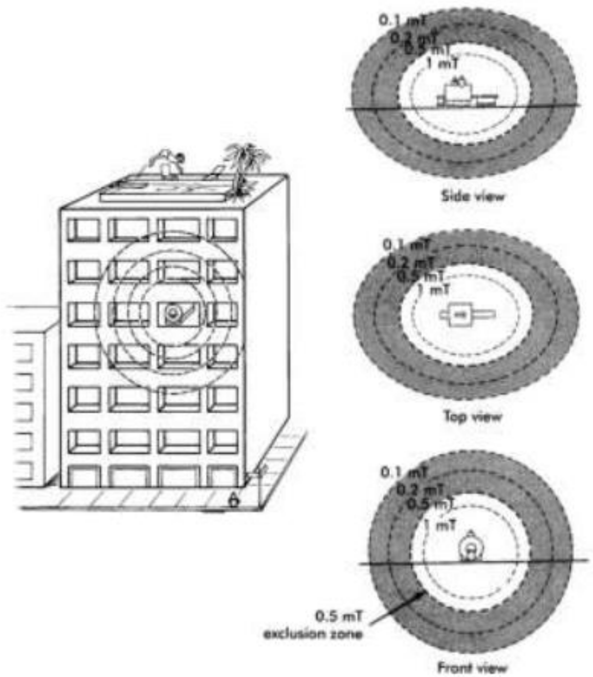
| Advantages | Disadvantages |
|---|---|
| **Permanent** | |
| Low capital cost | Limited field strength |
| Low operating cost | Fixed field strength |
| Negligible fringe field | Very heavy |
| **Resistive Iron Core** | |
| Low capital cost | High power consumption |
| Easy coil maintenance | Water 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 |

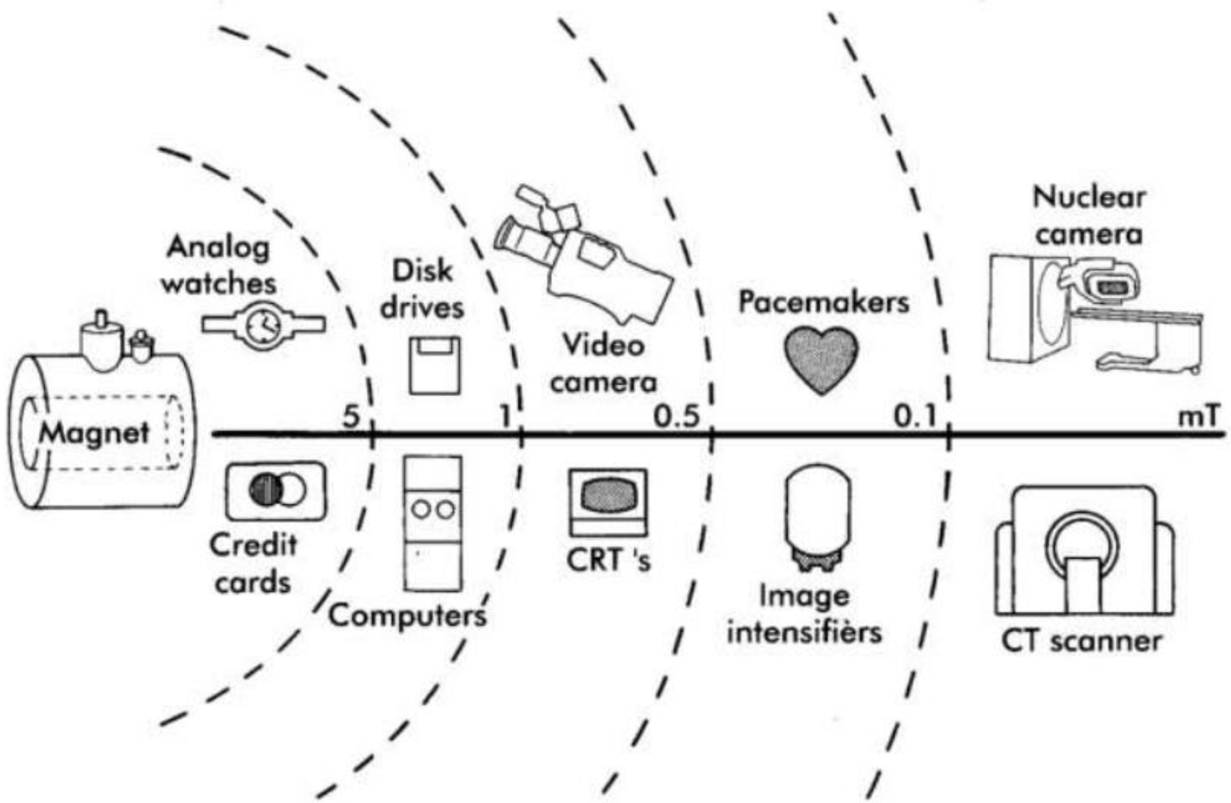
# Site Selection for MRI

**Table 13-3** Considerations for locating a magnetic resonance imager

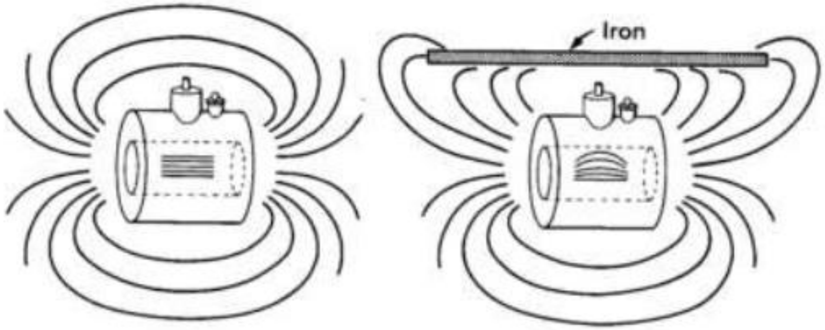
| Advantages | Disadvantages |
|---|---|
| **New construction** | |
| Easier to plan for fringe magnetic field | Cost |
| Custom design | Possibly remote |
| **Existing building** | |
| Proximity to other services | Accommodation of fringe magnetic field, higher renovation cost |
| Use of existing facilities | |
| **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 |

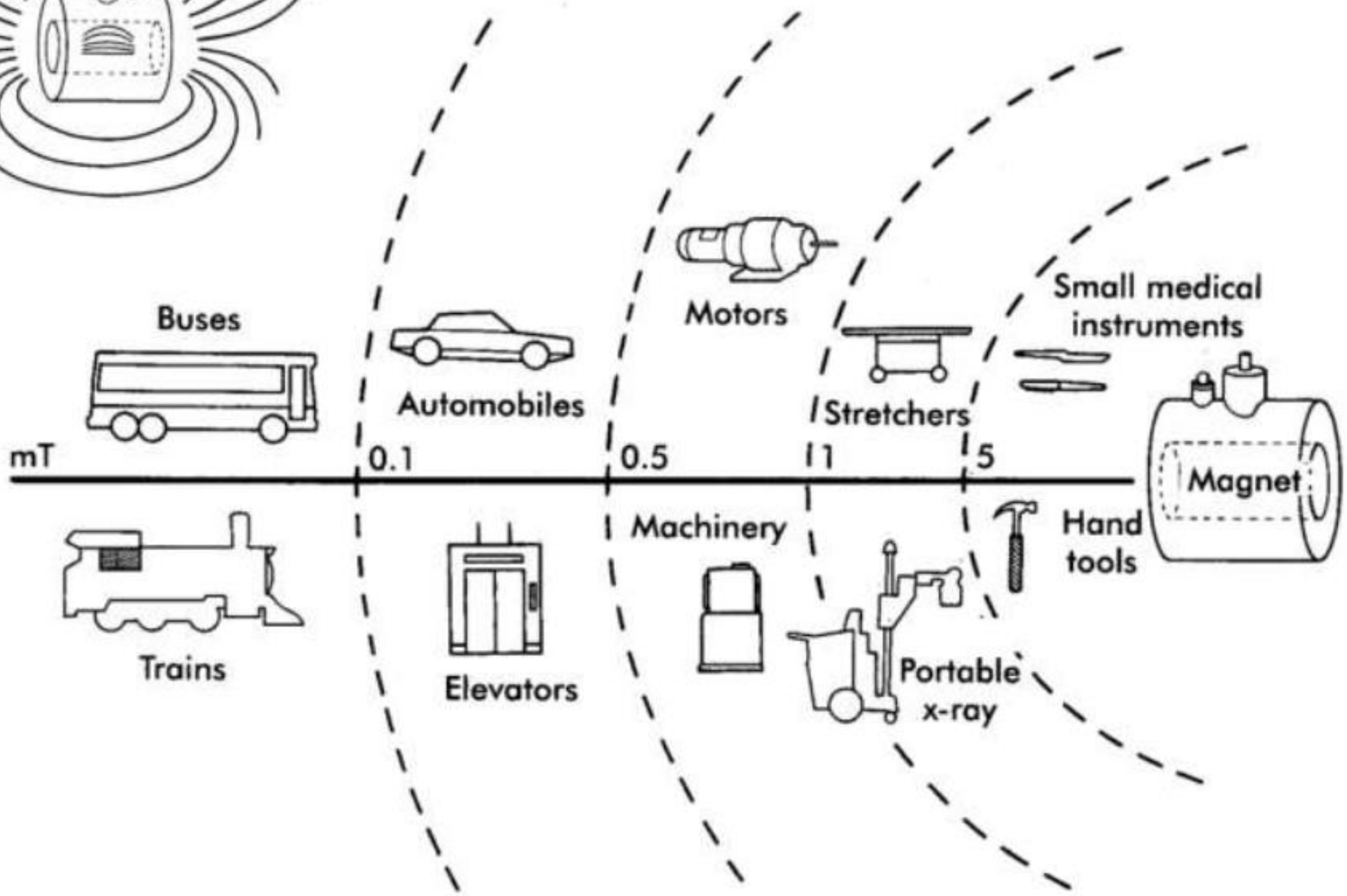
# Effects of MRI on the Environment



# Effects of the Environment on MRI



# Suggested Problem Sets

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