EL5823/BE6203 --- Medical Imaging - I

MRI Instrumentation and Pulse Sequences

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Based on J. L. Prince and J. M. Links, Medical Imaging Signals and Systems, and lecture notes by Prince. Figures are from the textbook except otherwise noted.

Lecture Outline

- Review of MRI physics and imaging principle
- MRI instrumentation
  - Magnet, gradient coil, RF coil
- Pulse sequences for slice selection and position encoding
  - Slice selection
  - Rectilinear scanning
  - Polar scanning
- Pulse parameter selection for T1,T2,PD weighting

Summary: Process Involved in MRI

- Put patient in a static field B_0 in z-direction
- (step 1) Wait until the bulk magnetization M reaches an equilibrium (M align with B_0)
- Apply a rotating field (alpha pulse) B_1 in the xy plane to bring M to an initial angle \alpha with B_0. Typically \alpha=\pi/2
- M precesses around B_0 (z direction) at Larmor freq. with angle \alpha
- The component in z increases in time (longitudinal relaxation) with time constant T1
- The component in x-y plane reduces in time (transverse relaxation) with time constant T2
- Apply \pi pulse to induce echos to bring transverse components in phase to increase signal strength
- Measure the transverse component at different times (NMR signal) (typically at echo time)
- Go back to step 1
- By using different excitation pulse sequences (differing in TE, TR, \alpha), the signal amplitude can reflect mainly the proton density, T1 or T2 at a given voxel

Unanswered Question

- How to measure the signal at one particular location?
  - Using gradient coils to yield a static field which changes in location (B_0(x,y.z)) and hence the Larmor freq. changes in x,y,z
  - Apply RF pulses in a certain range so that only a certain voxels are excited or measured
- Actually measure samples of the Fourier transform of a slice
  - Need reconstruction

MRI Scanner Components

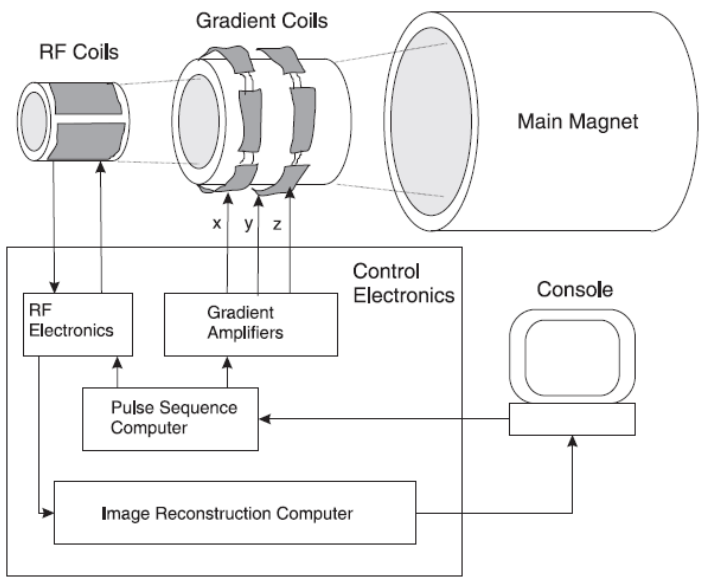
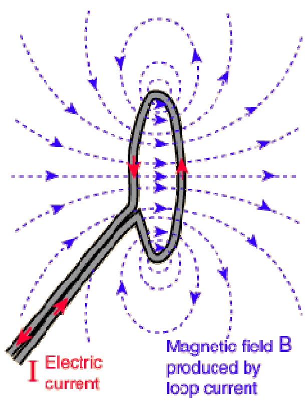


Image of a Practical MRI Machine

How to generate Magnetic field?

- From a current loop
- From a straight wire with current

Magnetic Field of a Current Loop

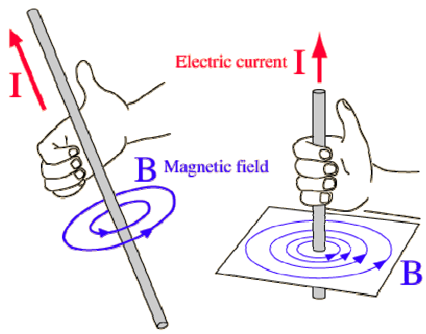


$$B \propto \frac{IR^2}{(z^2 + R^2)^{3/2}}$$

Center strength proportional to I/R
Decrease with distance (1/z^3)

From http://hyperphysics.phy-astr.gsu.edu/hbase/hframe.html

Magnetic Field of a Current Wire



$$B \propto \frac{I}{r} \text{(for infinitely long wire)}$$

From: http://hyperphysics.phy-astr.gsu.edu/hbase/hframe.html

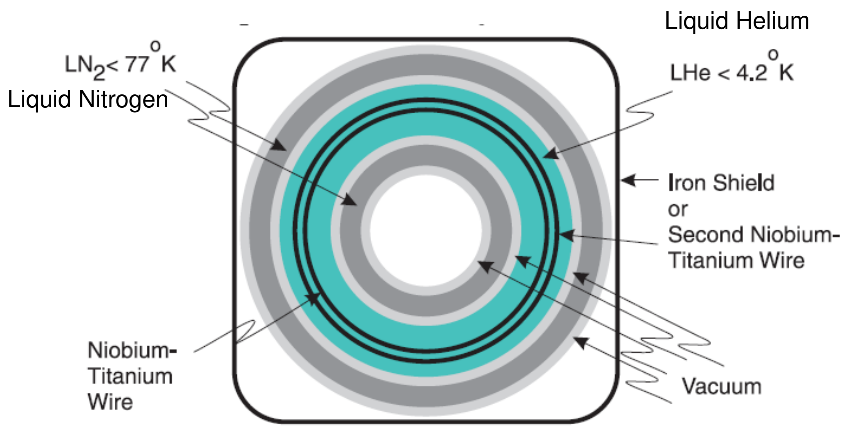
Magnet for Static Field B_0

- Demands:
  - Spatially homogeneous field
  - Stable over time
  - Strong field
  - Patient access (volume)
- Permanent magnet:
  - < 3% of all magnets
  - < 0.3 T
  - Economic, open
  - Small fringe field
- Resistive magnet
  - Using a current loop through a metal wire
  - ~ 50 kW, needs cooling water
  - Heat constrains maximum current -> 0.15 - 0.3 T

Superconducting magnet

- Superconducting magnet:
  - Using a superconductive Niobium/Titanium alloy (Type II, 1950) to carry current, supports high magnetic field strengths
  - Superconducting wire has a resistance approximately equal to zero when it is cooled to a temperature close to absolute zero (-273.15o C or 0 K) by immersing it in liquid helium. Once current is caused to flow in the coil it will continue to flow as long as the coil is kept at liquid helium temperatures.
  - Once current is caused to flow in the coil it will continue to flow as long as the coil is kept at liquid helium temperatures.
  - 200 A / 1-10 T (1.5 T most common)
  - Decay 0.05 ppm/hour → years of operation
  - Bore in 1 meter, length 2.6-2.8 m
- Tradeoffs:
  - High field strength → reduced T1 contrast
  - Reduced RF penetration due to higher frequencies
  - Cost
  - Safety: 5-G line (~ 10-12 m from unshielded 1.5 T-magnet)

Superconducting Magnet



The length of superconducting wire in the magnet is typically several miles. The coil of wire is kept at a temperature of 4.2K by immersing it in liquid helium. The coil and liquid helium is kept in a large dewar. The typical volume of liquid Helium in an MRI magnet is 1700 liters. This dewar is surrounded by a liquid nitrogen (77.4K) dewar which acts as a thermal buffer between the room temperature (293K) and the liquid helium.

Magnet Specification

- Field strengths from 0.5T to 9.0T
- Most common field strength: 1.5T
- Shimming to maintain homogeneous field
  - passive shimming
  - active shimming
  - better than ± 5 ppm required
- Minimize fringe field (outside the bore)
  - nuisance and dangerous
  - passive: iron shield, or
  - active: second superconducting wires

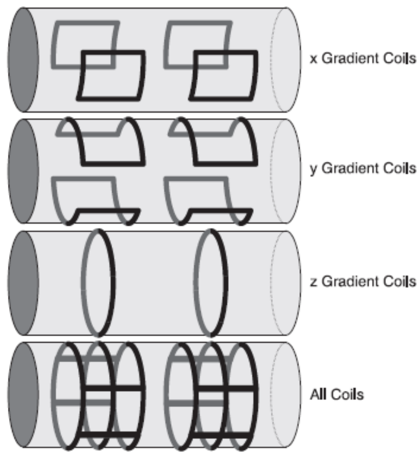
Gradient Coils

- Fit just inside the bore
- Role: change B_0 as a function of position
- Three coils:
  - x, y, and z directions
  - G_x, G_y, and G_z strengths
- Modify main field as follows

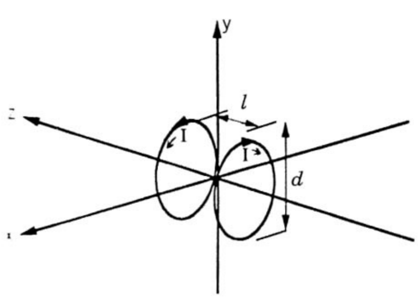
$$\mathbf{B} = (B_0 + G_x x + G_y y + G_z z)\hat{z}$$

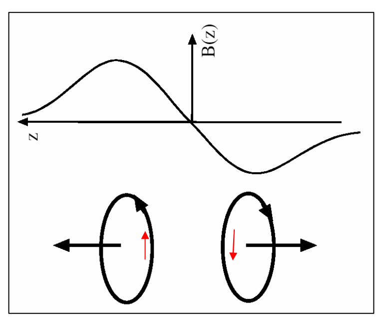
- This is the key to MR imaging

- x and y are saddle coils
- z is opposing coils

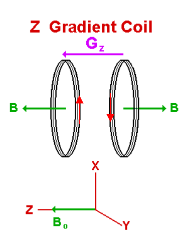


Z-Gradient





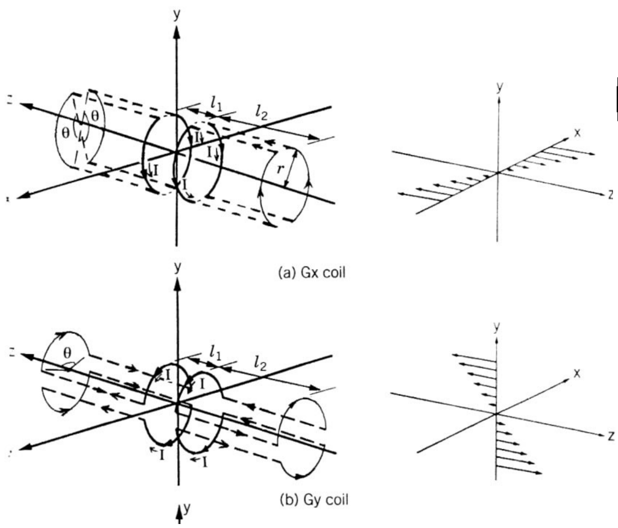
Graber, lecture note for BMI F05



antihelmholtz coil

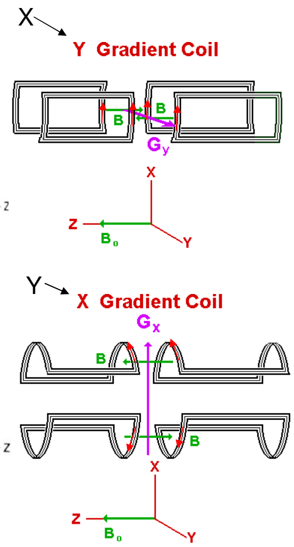
http://www.cis.rit.edu/htbooks/mri

X- and Y-Gradient



Graber, lecture note for BMI F05

Saddle coil



http://www.cis.rit.edu/htbooks/mri

Specification of Gradient Coils

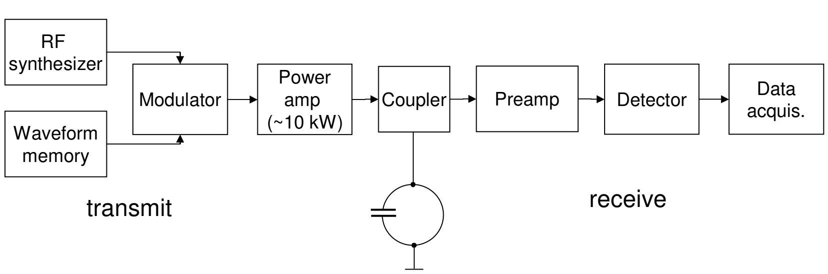
- Maximum gradient 1–6 Gauss/cm
- <u>Switching times</u> 0.1–1.0 ms
- <u>slew rates</u> 5–250 mT/m/msec
- Additional shielding outside to reduce eddy currents
- FDA limit 40 T/s

RF Coils

- RF coils create the B1 field which rotates the net magnetization in a pulse sequence. (transmission mode)
- They also detect the transverse magnetization as it precesses in the XY plane. (receive mode)
- Three general categories; 1) transmit and receive coils, 2) receive only coils, and 3) transmit only coils. Near-field antennas
- Coils are resonant circuits, tuned w/ capacitors for efficient transmitting and receiving at Larmor frequency (improved SNR)
  - w0=1/sqrt(LC)
- Safety: limit absorbed power to prevent heating in excess of 1℃

RF transmitter and receiver

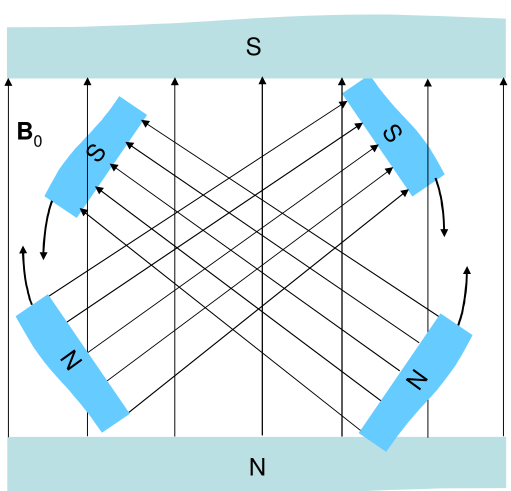
- Specific waveforms (truncated sinc pulse) are achieved by digital waveform memory (Bandwidth of slice selection ~ 20 kHz)
- Carrier from RF synthesizer modulated by waveform
- Power amp (~ 10 kW)
- Receiving preamp (100 kHz), detector, data acquisition, storage



How does RF Coil Generate Rotating Magnetic Field

- The two counter rotating magnetic fields in x-y plane produces a Sinusoidal Field in x-direction with the freq = the rotation frequency (=Larmor freq.)
- An RF coil generates a sinusoidal field in x-direction

Desired Rotating Magnetic Fields

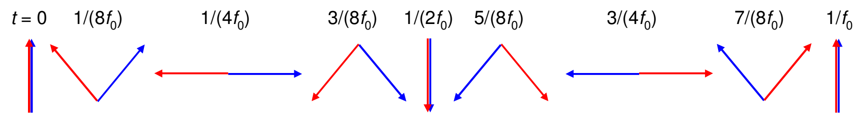


Two magnets, whose fields are ϒ $\mathbf{B}_0$, that <u>rotate</u>, in opposite directions, at the Larmor frequency

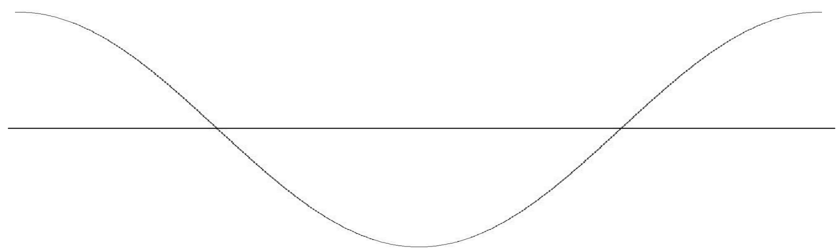
From Graber, lecture note for BME F05

Resulting Magnetic Field in x-Direction

Simplified bird's-eye view of counter-rotating magnetic field vectors

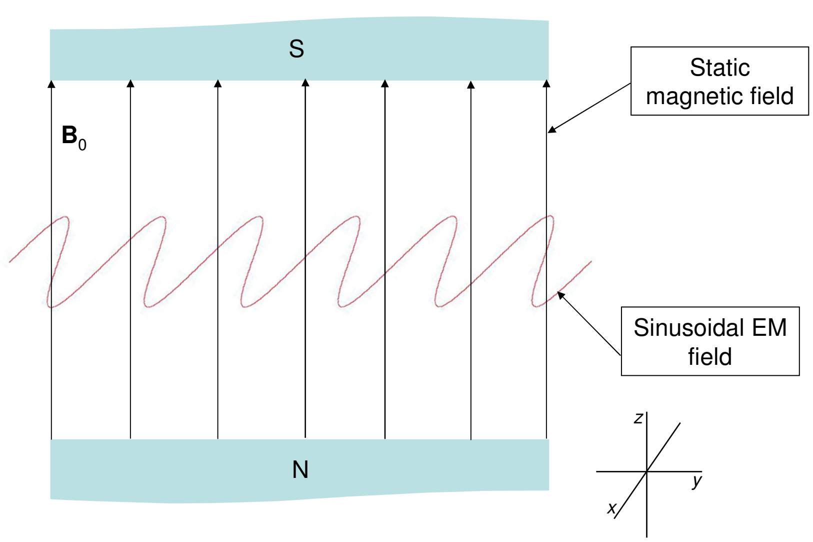


So what does resulting **B** *vs. t* look like?



This time-dependent field is called $\mathbf{B}_1$.

From Graber, lecture note for BME F05



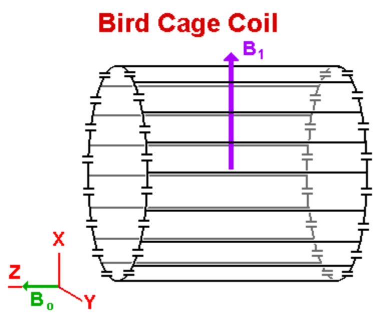
From Graber, lecture note for BME F05

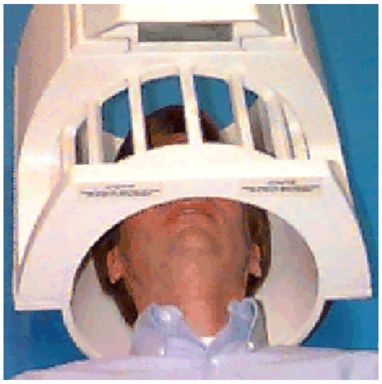
Types of RF Coils

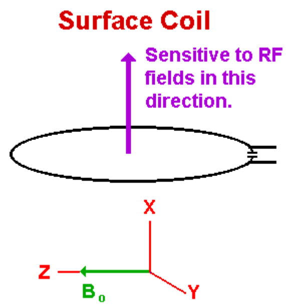
- Coil types
  - Homogeneous field coils (Head, whole body)
    - Birdcage
    - Saddle
  - Surface coils (local anatomy, e.g. spine), often receive-only coils
  - Different coils may be used to imaging different body parts
- Operate at frequencies in the range of ~1-170 MHz

Different Types of RF Coils

- See http://www.cis.rit.edu/htbooks/mri/inside.htm (chap 9: imaging hardware)







http://www.cis.rit.edu/htbooks/mri/inside.htm

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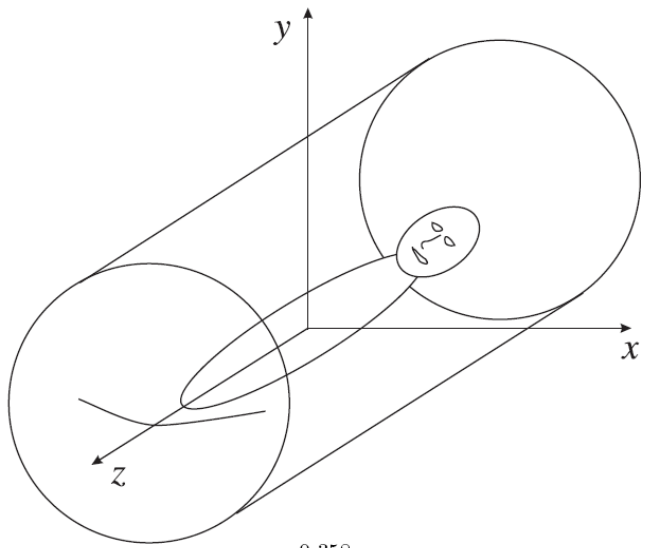
Laboratory Coordinate



Larmor Frequency Encoding Using Gradient Fields

- Gradient $\mathbf{G} = (G_x, G_y, G_z)$ produces B-field:

$$\mathbf{B} = (B_0 + \mathbf{G} \cdot \mathbf{r})\hat{z}$$

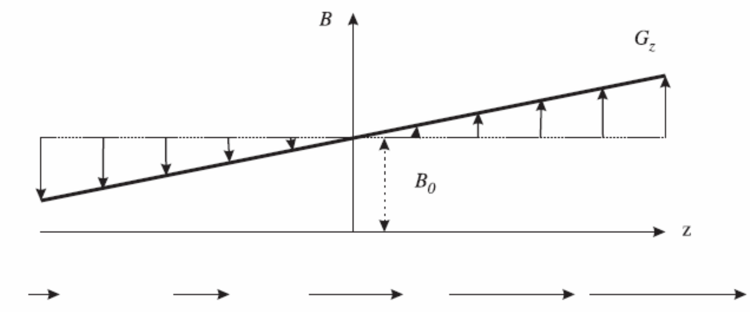
  where $\mathbf{r} = (x, y, z)$
- Spatially varying Larmor frequency

$$\nu(\mathbf{r}) = \gamma(B_0 + \mathbf{G} \cdot \mathbf{r})$$

Slice Selection Using Z-Gradient

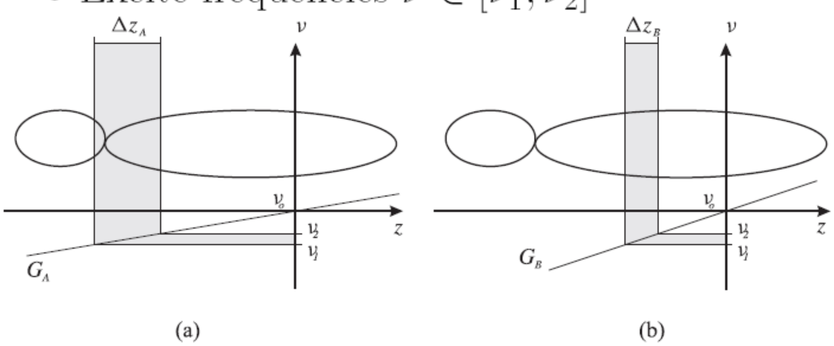
- Let $\mathbf{G} = (0, 0, G_z)$
- Then

$$\nu(\mathbf{r}) = \nu(z) = \gamma(B_0 + G_z z)$$



Slice Excitation

- Excite frequencies $\nu \in [\nu_1, \nu_2]$



- Causes "slab" excitation of spin system

Slice Selection Parameters

- RF parameters:

$$\bar{\nu} = \frac{\nu_1 + \nu_2}{2} \quad \text{center frequency}$$

$$\Delta\nu = |\nu_2 - \nu_1| \quad \text{frequency range}$$

- Slice parameters:

$$\bar{z} = \frac{\bar{\nu} - \nu_0}{\gamma G_z} \quad \text{slice position}$$

$$\Delta z = \frac{\Delta\nu}{\gamma G_z} \quad \text{slice thickness}$$

"Ideal" RF Excitation Pulse

- Excite frequencies in range $[\nu_1, \nu_2]$ Hz
- Excitation signal has Fourier transform

$$S(\nu) = A \text{ rect}\left(\frac{\nu - \bar{\nu}}{\Delta\nu}\right)$$

- Signal is $s(t) = A\Delta\nu \text{ sinc}(\Delta\nu t)e^{j2\pi\bar{\nu}t}$

Tip angle is $$\alpha(z) = \gamma A\tau_p \text{rect}\left(\frac{z - \bar{z}}{\Delta z}\right)$$

Derivation of Tip Angle

- Excitation is at freq. w0, denoted by $B_1(t) = A\Delta v \operatorname{sinc}(\Delta v t) e^{j\omega_0 t}$
- Recall that tip angle is related to B_1^e(t) by $\alpha = \gamma \int_{-\infty}^{\infty} B_1^e(t)\,dt$
- When the spin Larmor freq. w is not the same as the excitation w0, we need to replace B_1^e(t) by

$$B_1^e(\omega, t) = B_1(t) e^{-j\omega t} = A\Delta v \operatorname{sinc}(\Delta v t) e^{j(\omega_0 - \omega)t}$$

- The Larmor freq. at z is $\omega(z) = \gamma(B_0 + G_z z); \omega_0 = \omega(\bar{z}) = \gamma(B_0 + G_z \bar{z})$
- Show on the board: $\alpha(z) = \gamma A \operatorname{rect}\left(\frac{z - \bar{z}}{\Delta z}\right)$
- Note that \alpha(z) has the same profile as S(v). The slice thickness can be translated from S(v) bandwdith using $\Delta z = \frac{\Delta v}{\gamma G_z}$

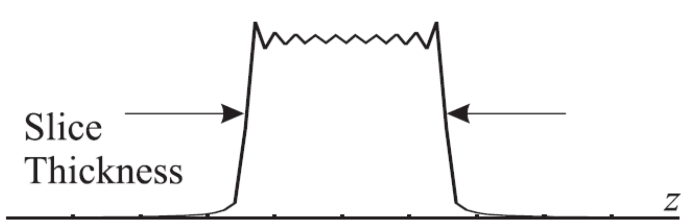
Practical RF Excitation Pulse

- Truncated sinc

$$\tilde{s}(t) = \left[A\Delta\nu \operatorname{sinc}(\Delta\nu t) e^{j2\pi\bar{\nu}t}\right] \operatorname{rect}(t/\tau_p)$$

- Corresponding tip angle profile:

$$\alpha(z) = \gamma A \tau_p \operatorname{rect}\left(\frac{z - \bar{z}}{\Delta z}\right) * \operatorname{sinc}\left(\tau_p \gamma G_z (z - \bar{z})\right)$$



Slice Dephasing and Refocusing

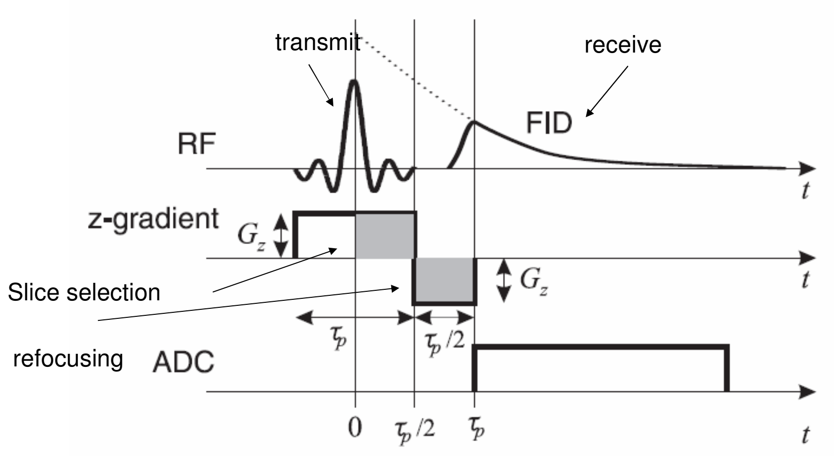
- Difference Larmor frequencies across slice:
  - "slow" on low side
  - "fast" on high side
- Phase difference is

$$\phi(z) = \gamma G_z (z - \bar{z}) \tau_p / 2$$

- Refocus with negative gradient pulse
  - strength $-G_z$
  - duration $\tau/2$

Why \tau_p/2?
Spins start to precess coherently roughly at \tau_p/2

A Simple Pulse Sequence



Basic Signal Model

- The transverse magnetic field M_xy at all voxels in a slice induces a voltage signal (FID) in the surrounding RF coil (receive mode)
- FID depends on the integration of the magnetic field over all voxels

$$s(t) = \iint_{slice} M_{xy}(x, y, t)\,dxdy$$

- M_xy follows transverse relaxation after excitation:

$$M_{xy}(x, y, t) = M(x, y, 0^+)e^{i\omega_0 t}e^{-t/T_2}$$

$$M(x, y, 0^+) = M_0(x, y)\sin\alpha$$

$$M_0(x, y) = \frac{B_0\gamma^2 h^2}{4\kappa T}P_D(x, y)$$

Note that M(t) follows T2* decay unless echoes are induced.

When we measure FID, T2 in above equation should be replaced by T2*.

When we measure echoes (using spin-echo pulse sequences), we use T2.

Measured Signal:

$$s(t) = e^{j\omega_0 t}\iint_{slice} M(x, y, 0^+)e^{-t/T_2}dxdy = s_0(t)e^{j\omega_0 t}$$

Baseband signal (obtained after demodulation)

$$s_0(t) = \iint_{slice} f(x, y; t)dxdy$$

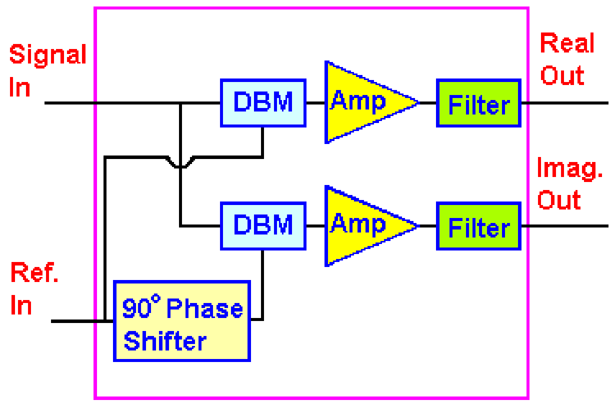
Effective Spin Density:

$$f(x, y; t) = M(x, y, 0^+)e^{-t/T_2}$$
$$\approx M(x, y, 0^+) \quad \text{when } t << T_2$$

Recall $M(x, y, 0^+)$ is proportional to spin density $P_D(x, y)$.

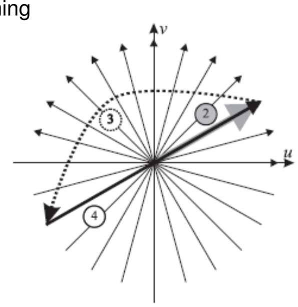
MRI images are images of *f(x,y,t)*, which is mainly inflenced by $P_D(x, y)$ at $t << T2$.

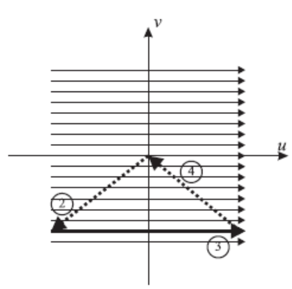
Demodulation of the Received Signal



Locating a Voxel in a Slice

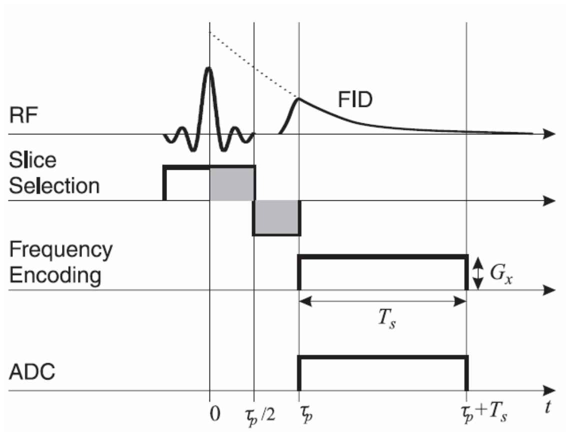
- The FID gives the sum of signals from all voxels in a slice
- How do we separate contributions from different voxels?
- Apply x- and y-direction gradient fields so that voxels in different x,y positions have different Larmor freq.
- Instead of scanning the spatial position (x,y), scan in 2D frequency space (u,x)
  - Polar scanning
  - Rectilinear scanning





X-Position Encoding (frequency encoding)

- Apply a Gx field over a certain time period Ts, measure the signal during this time (ADC window)



Question: can we apply the Gx gradient with the z-gradient simultaneously?

Signal Model

- Larmor frequency is function of x

$$\nu(x) = \gamma(B_0 + G_x x)$$

- Baseband signal becomes

$$s_0(t) = \int_{-\infty}^{\infty}\int_{-\infty}^{\infty} f(x,y)e^{-j2\pi\gamma G_x x t}\,dx\,dy$$

Relation to 2D FT

- Recall 2D FT: $F(u,v) = \iint f(x,y) e^{-j2\pi(ux+vy)} dxdy$
- Our baseband signal:

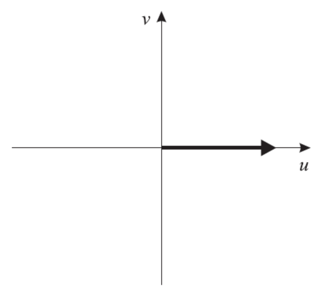
$$s_0(t) = \int_{-\infty}^{\infty}\int_{-\infty}^{\infty} f(x,y) e^{-j2\pi\gamma G_x x t}\, dx\, dy$$

$s_0(t) = F(u = \gamma G_x t, v = 0)$

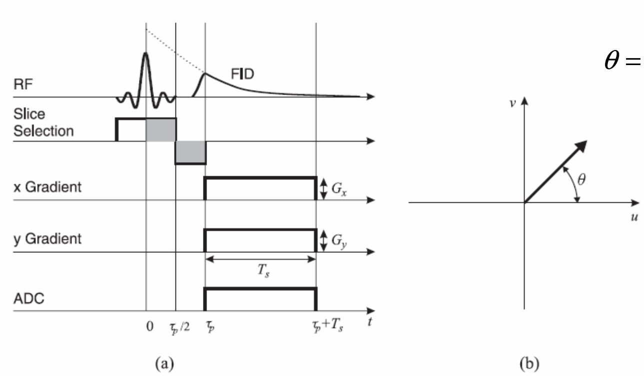
$F(u,0) = s_0(u/\gamma G_x)$

Samples taken over $t \in (0, T_s)$

Corresponding to freq. $u \in (0, \gamma G_x T_s$



Polar Scanning



$\theta = \tan^{-1}\frac{G_y}{G_x}$

$u = \gamma G_x t \qquad v = \gamma G_y t$

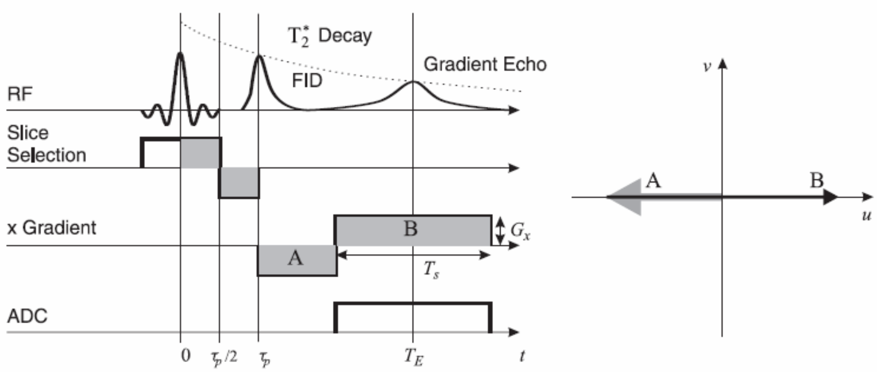
Repeat above pulses after the magnetization returns to equilibrium.
In each new cycle, change Gx,Gy to scan a different angle

Rectilinear Scanning

- Need to move u to negative axis (gradient echo)
- Need to vary v position in different cycles (phase encoding)

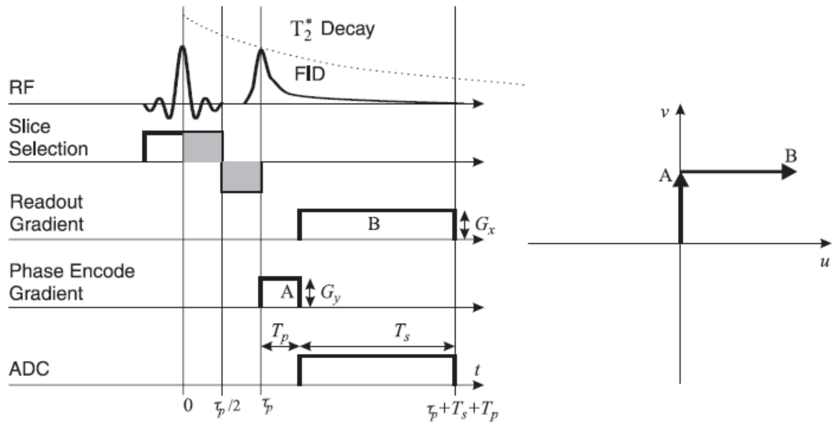
Gradient Echo

- Moving u to negative starting point

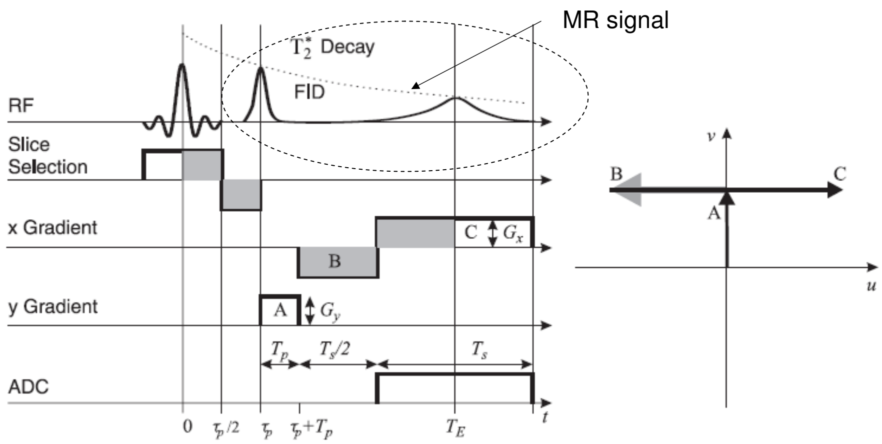


Phase Encoding Sequence

- Moving scan to a different line (v) through Y-gradient



Gradient Echo Pulse Sequence



Phase encode signal model

- Accumulated phase after phase encode

$$\phi_y(y) = -\gamma G_y T_p y$$

- Baseband signal during readout

$$s_0(t) = \int_{-\infty}^{\infty}\int_{-\infty}^{\infty} f(x, y) e^{-j\gamma G_x x t} e^{-j\gamma G_y T_p y} \, dx \, dy$$

- Recognize Fourier transform frequencies:

$$\begin{aligned} u &= \gamma G_x t \\ v &= \gamma G_y T_p \end{aligned}$$

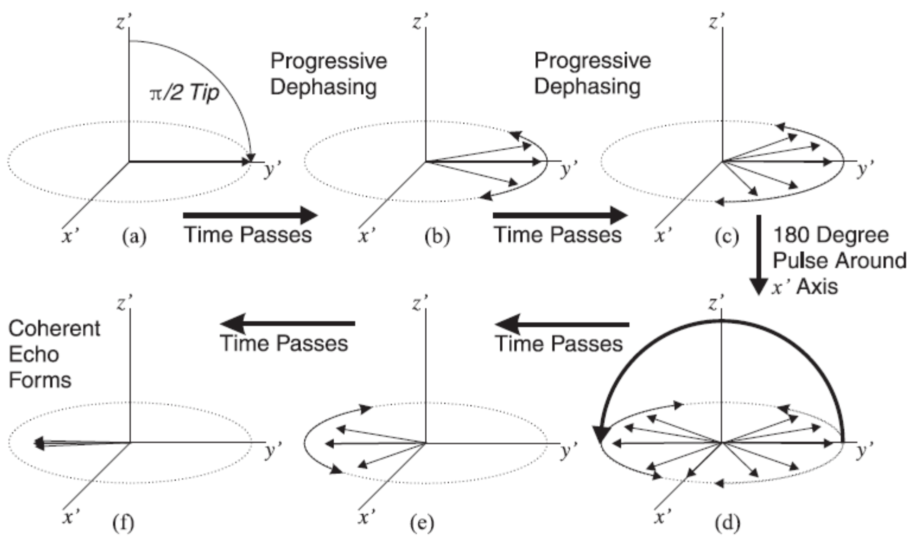
Spin Echo Measurement

- Previous pulse sequences are used to measure the FID, not the spin echo.
- The signal experiences T2^* decay
- To measure spin echo (which follows T2 decay), need to apply inverse pulse before measurement

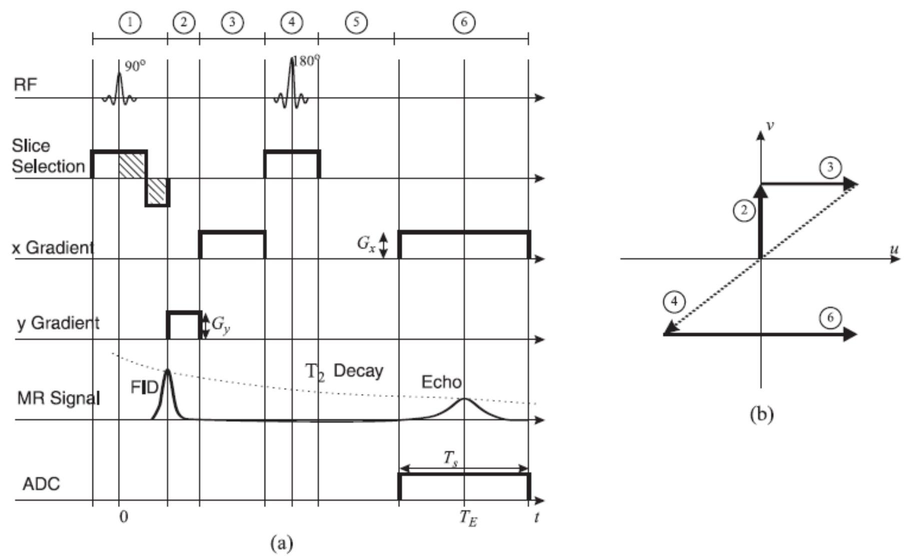
- See animation at
- http://www.cis.rit.edu/htbooks/mri/inside.htm

- Chap 7, section on FT tomographic imaging

Spin Echo (review)



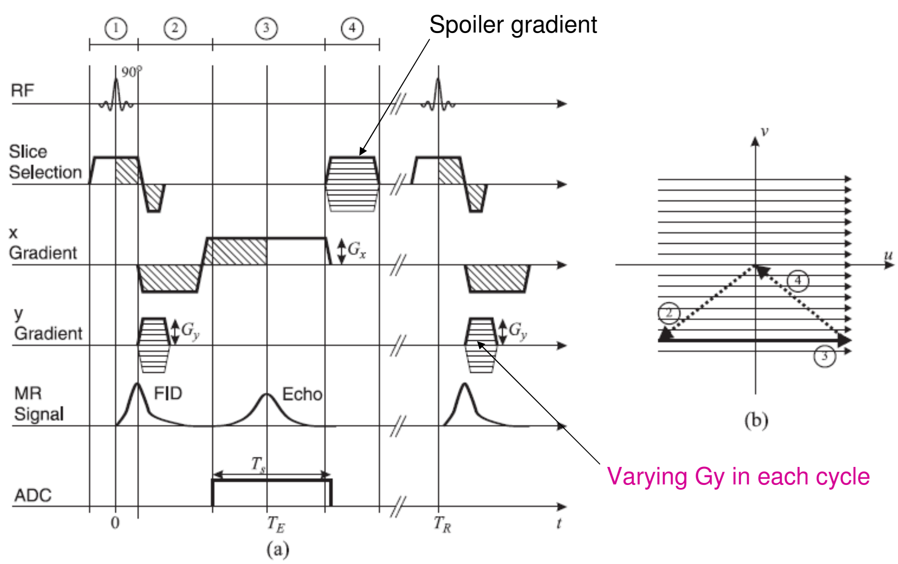
Spin-Echo Sequence



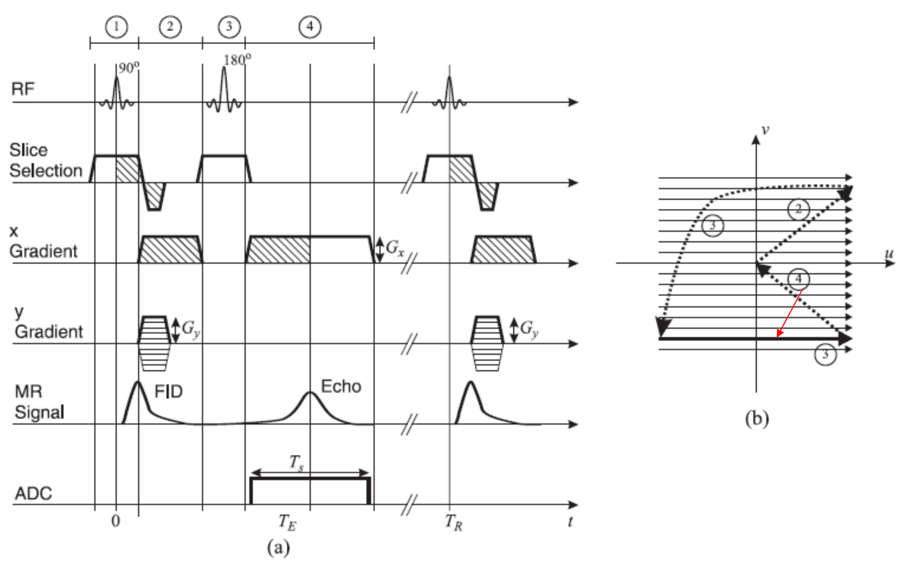
Pulse Repetition

- To scan different lines in the frequency space, we need to repeat the previous steps with a different Gy (rectilinear scan) or different Gx,Gy pair (polar scan)
- But we should wait a period
  - Slow imaging sequence
    - T_R>> T_2
    - (M_xy has completely disappeared before the next RF pulse)
  - Fast imaging sequence
    - Use multiple spin echos after a single RF excitation
    - Spoiling Mxy prior subsequent excitation
      - By appling a z-gradient field, dephase spins in the selected slice so that spins added over the entire slice add destructively and hence no MR signal will be produced until the next excitation

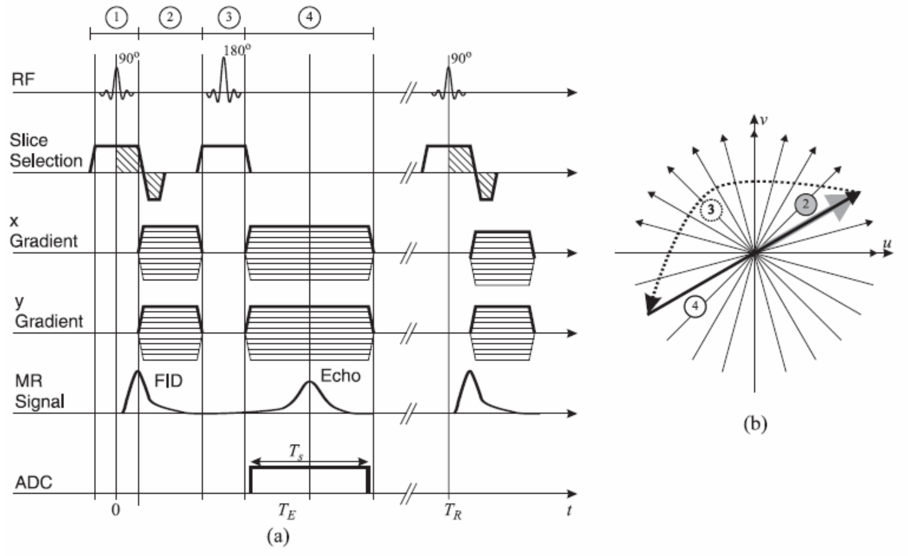
Realistic Gradient Echo Pulse Sequence



Realistic Spin Echo Pulse Sequence



Realistic Spin-Echo Polar Pulse Sequence



Steady State Response

Recall $f(x,y,t) = AM(x,y;0^+)e^{-t/T_2(x,y)}$

If T_R is sufficiently long ($>> T_1$), $M(x,y;0^+) = M_0(x,y)\sin\alpha$

What if T_R is shorter ($\approx T$)? (to reduce scan time)

As long as $T_R >> T_2$, after many pulse cycles, each spin reaches a steady state $M_z \neq M_0$.

Let this steady state value be $M_z^\infty(x,y)$, then $M(x,y;0^+) = M_z^\infty(x,y)\sin\alpha$

We can show $M_z^\infty = M_0(x,y)\dfrac{1-e^{-T_R/T_1}}{1-\cos\alpha\, e^{-T_R/T_1}}$ (see proof next page)

Therefore

$$f(x,y,t) = AM_0(x,y)\sin\alpha\, e^{-t/T_2(x,y)}\frac{1-e^{-T_R/T_1}}{1-\cos\alpha\, e^{-T_R/T_1}}$$

$$\boxed{\propto P_D(x,y)\sin\alpha\, e^{-t/T_2(x,y)}\frac{1-e^{-T_R/T_1}}{1-\cos\alpha\, e^{-T_R/T_1}}}$$

When $\alpha = \pi/2$

$$\boxed{f(x,y,t) \propto P_D(x,y)e^{-t/T_2(x,y)}\left(1-e^{-T_R/T_1}\right)}$$

This is the imaging equation for MRI.

In spin-echo sequence, we measure at $T_E << T_2$

Proof

Recall the longitudinal relaxation after the n - th alpha pulse follows :

$M_{z,n}(t) = M_{z,n-1}(0+)\cos\alpha e^{-t/T_1} + M_0(1-e^{-t/T_1})$

where $M_{z,n-1}(0+)$ is the M_z at the end of the (n - 1) - th pulse sequence.

Before first pulse $M_{z,0}(0+) = M_0$, M_z at time TR after first pulse is

$M_{z,1}(T_R) = M_0 \cos\alpha e^{-T_R/T_1} + M_0(1-e^{-T_R/T_1})$

Therefore, just second pulse, $M_{z,1}(0+) = M_{z,1}(T_R)$.

M_z at time TR after second pulse is

$M_{z,2}(T_R) = M_{z,1}(0+)\cos\alpha e^{-T_R/T_1} + M_0(1-e^{-T_R/T_1})$

$= M_{z,1}(T_R)\cos\alpha e^{-T_R/T_1} + M_0(1-e^{-T_R/T_1})$

Continue above, we have

$M_{z,n}(T_R) = M_{z,n-1}(0+)\cos\alpha e^{-T_R/T_1} + M_0(1-e^{-T_R/T_1})$

$= M_{z,n-1}(T_R)\cos\alpha e^{-T_R/T_1} + M_0(1-e^{-T_R/T_1})$

At steady state

$M_{z,n}(T_R) = M_{z,n-1}(T_R) = M_z^\infty$

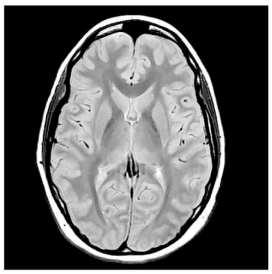
$M_z^\infty = M_z^\infty \cos\alpha e^{-T_R/T_1} + M_0(1-e^{-T_R/T_1})$

$$M_z^\infty = M_0 \frac{1-e^{-T_R/T_1}}{1-\cos\alpha\, e^{-T_R/T_1}}$$

T1,T2,PD Weighting - Revisit

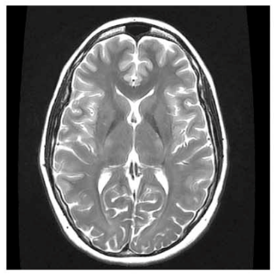
- Different tissues vary in T1, T2 and PD (proton density)
- The pulse sequence parameters can be designed so that the captured signal magnitude is mainly influenced by one of these parameters
- Pulse sequence parameters
  - Tip angle \alpha
  - Echo time T_E
  - Pulse repetition time T_R

	P_D	T_2 (ms)	T_1 (ms)
White matter	0.61	67	510
Gray matter	0.69	77	760
CSF	1.00	280	2650



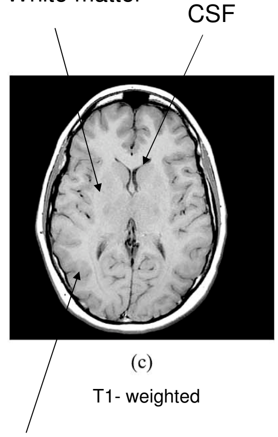
(a)

PD weighted



(b)

T2- weighted



(c)

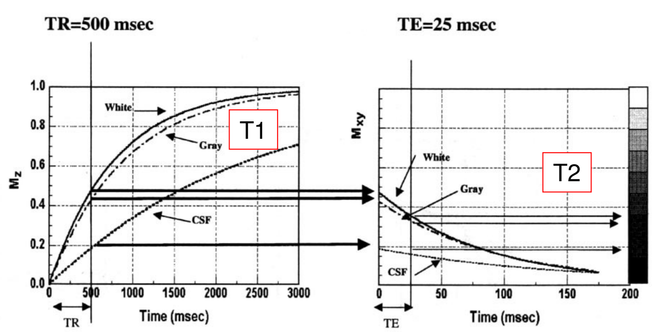
T1- weighted

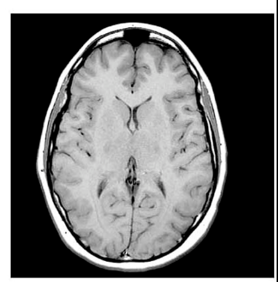
T1-weighting

- Short TR:
  - Maximizes T1 contrast due to different degrees of saturation
  - If TR too long, tissues with different T1 all return equilibrium already
- Short TE:
  - Minimizes T2 influence, maximizes signal

$$f(x,y,T_E) \propto P_D(x,y)\, e^{-T_E/T_2(x,y)}\left(1-e^{-T_R/T_1}\right)$$

Note that we measure only M_xy, but TR influences the starting position of M_xy (initial position of T2 relaxation)

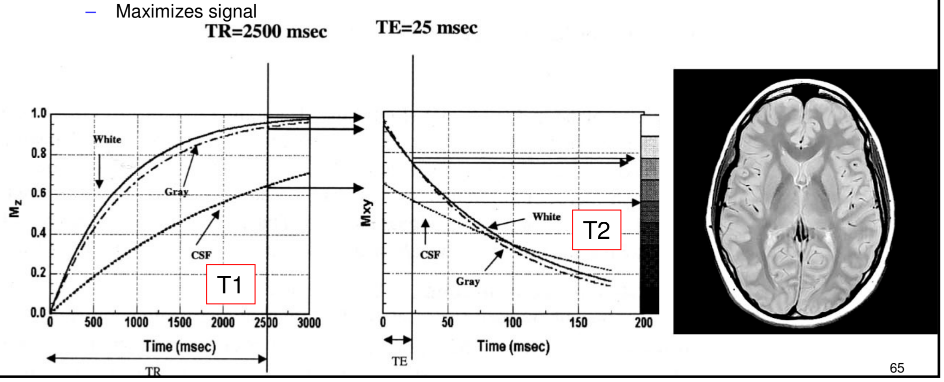




Spin density weighting

- Signal proportional to PD
- Long TR:
  - Minimizes effects of different degrees of saturation (T1 contrast)
  - Maximizes signal (all return to equilibrium)
- Short TE:
  - Minimizes T2 contrast
  - Maximizes signal

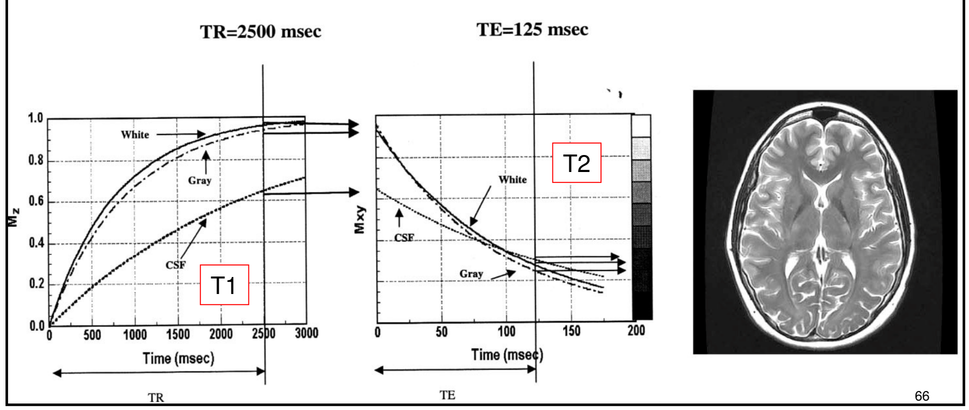
$$f(x,y,T_E) \propto P_D(x,y)\, e^{-T_E/T_2(x,y)}\left(1-e^{-T_R/T_1}\right)$$



T2 weighting

- Long TR:
  - Minimizes influence of different T1
- Long TE:
  - Maximizes T2 contrast
  - Relatively poor SNR

$$f(x,y,T_E) \propto P_D(x,y)\, e^{-T_E/T_2(x,y)}\left(1-e^{-T_R/T_1}\right)$$



Example

- Example 13.9
- Suppose two tissues have same PD, T2, but different T1. How should you choose TR to maximizes the contrast between the two tissue in the recovered effective spin image. Assuming \alpha=pi/2 and we measure at TE using a spin echo pulse sequence.
- Go through on the board

Summary

- Design and functions of magnets, gradient coils and RF coils
- RF and gradient pulse sequences
- Slice selection using z-gradient
- Scanning in frequency domain in one slice
  - Rectilinear scan:
  - Polar scan:
  - FID measurement (using gradient echo sequence), follows T2* decay
  - Spin echo measurement (using spin echo sequence), measure at echo time, follows T2 decay
  - Should understand the purpose of each pulse in a given pulse sequence
    - Understand the relation between the pulse sequence and the trajectory on the frequency domain!
  - Difference between ideal and practical pulse sequences
- Should know relation between f(x,y) and s0(t) for different pulse sequences
  - Basis for image formation and reconstruction
- Relation between f(x,y) and the tissue properties (MR imaging equation)
  - How to vary pulse sequence parameters to weight T1, T2, PD contrast

Reference

- Prince and Links, Medical Imaging Signals and Systems, Chap. 13
- A. Webb, Introduction to Biomedical Imaging, Chap. 4
- **The Basics of MRI**, A web book by Joseph P. Horn (containing useful animation):
- http://www.cis.rit.edu/htbooks/mri/inside.htm
- C.E. Hayes, W.A. Edelstein, J.F. Schenck "Radio Frequency Resonators." In *Magnetic Resonance Imaging,* ed. by C.L. Partain, R.R. Price, J.A. Patton, M.V. Kulkarni, A.E. James Saunders, Philadelphia, 1988

Homework

- Reading:
  - Prince and Links, Medical Imaging Signals and Systems, Chap. 13 (sec. 13.1-13.3)
  - Note down all the corrections for Ch. 13 on your copy of the textbook based on the errata (see Course website or book website for update).
- Problems:
  - P13.2
  - P13.3
  - P13.4 (except part (d))
  - P13.12
  - P13.13