Imaging Sequences I

M219 - Principles and Applications of MRI

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2/21/2024

Course Overview

- 2024 course schedule
  - https://mrrl.ucla.edu/pages/m219_2024
- Assignments
  - Homework #3 is due on 3/10
- TA office hours, Weds 4-6pm
- Office hours, Fridays 10-12pm

RF Pulse Bandwidth and Slice Profile:
Small Tip Angle Approximation

Bloch Equation (at on-resonance)

$$\frac{d\vec{M}_{rot}}{dt} = \vec{M}_{rot} \times \gamma \vec{B}_{eff}$$

where $\vec{B}_{eff} = \begin{pmatrix} B_1(t) \\ 0 \\ \cancel{B_0 - \frac{\omega}{\gamma}} + G_z z \end{pmatrix}$

When we simplify the cross product,

$$\frac{d\vec{M}}{dt} = \begin{pmatrix} 0 & \omega(z) & 0 \\ -\omega(z) & 0 & \omega_1(t) \\ 0 & -\omega_1(t) & 0 \end{pmatrix} \vec{M}$$

$$\omega(z) = \gamma G_z z \qquad \omega_1(t) = \gamma B_1(t)$$

Small Tip Approximation

$$\frac{d\vec{M}}{dt} = \begin{pmatrix} 0 & \omega(z) & 0 \\ -\omega(z) & 0 & \omega_1(t) \\ 0 & -\omega_1(t) & 0 \end{pmatrix} \vec{M}$$

$M_z \approx M_0$ small tip-angle approximation

sin θ ≈ θ
cos θ ≈ 1
$M_z \approx M_0 \rightarrow$ constant

$$\frac{dM_z}{dt} = 0$$

$$\frac{dM_{xy}}{dt} = -i\gamma G_z z M_{xy} + i\gamma B_1(t) M_0$$

$$M_{xy} = M_x + iM_y$$

First order linear differential equation. Easily solved.

$$\frac{dM_{xy}}{dt} = -i\gamma G_z z M_{xy} + i\gamma B_1(t) M_0$$

Solving a first order linear differential equation:

$$M_{xy}(t,z) = i\gamma M_0 \int_0^t B_1(s) e^{-i\gamma G_z z\cdot(t-s)} ds$$

$$M_r(\tau, z) = iM_0 e^{-i\omega(z)\tau/2} \cdot \mathcal{FT}_{1D}\{\omega_1(t + \frac{\mathcal{T}}{2})\} \,|_{f=-(\gamma/2\pi)G_z z}$$

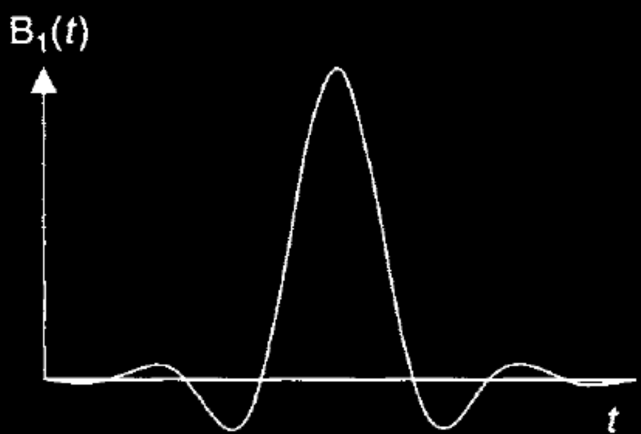
(See the note for complete derivation)

$$M_r(\tau, z) = iM_0 e^{-i\omega(z)\tau/2} \cdot \mathcal{FT}_{1D}\{\omega_1(t + \frac{\tau}{2})\} \mid_{f=-(\gamma/2\pi)G_z z}$$

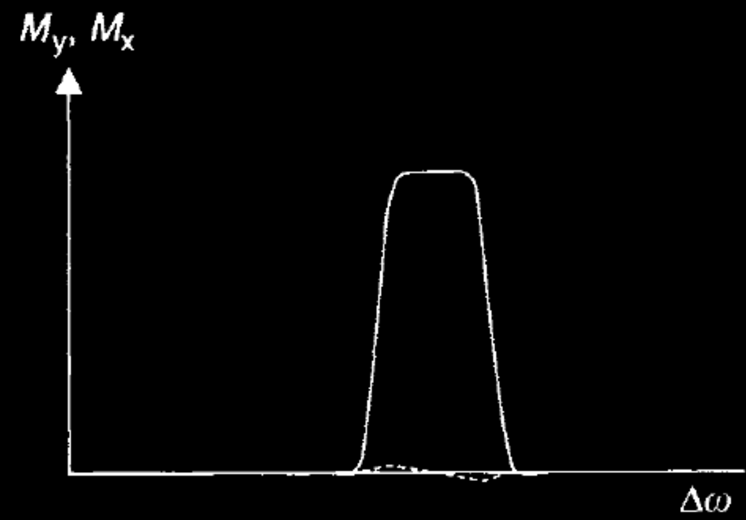
To the Board

Small Tip Approximation

$$M_r(\tau, z) = iM_0 e^{-i\omega(z)\tau/2} \cdot \mathcal{FT}_{1D}\{\omega_1(t + \frac{\mathcal{T}}{2})\} \,|_{f=-(\gamma/2\pi)G_z z}$$



Fourier transform



- For small tip angles, “the slice or frequency profile is well approximated by the Fourier transform of B1(t)”
- The approximation works surprisingly well even for flip angles up to 90°

Small Tip Approximation

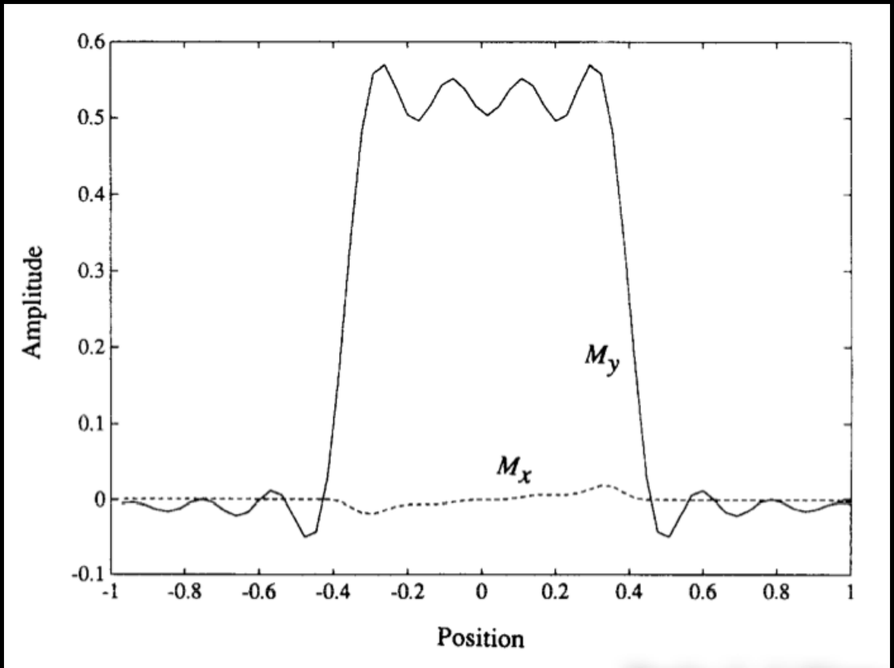
the excitation profile, within the small angle approximation, is just the Fourier transform of the pulse

remember that the Bloch equations are non-linear and thus cannot be expected to behave linearly

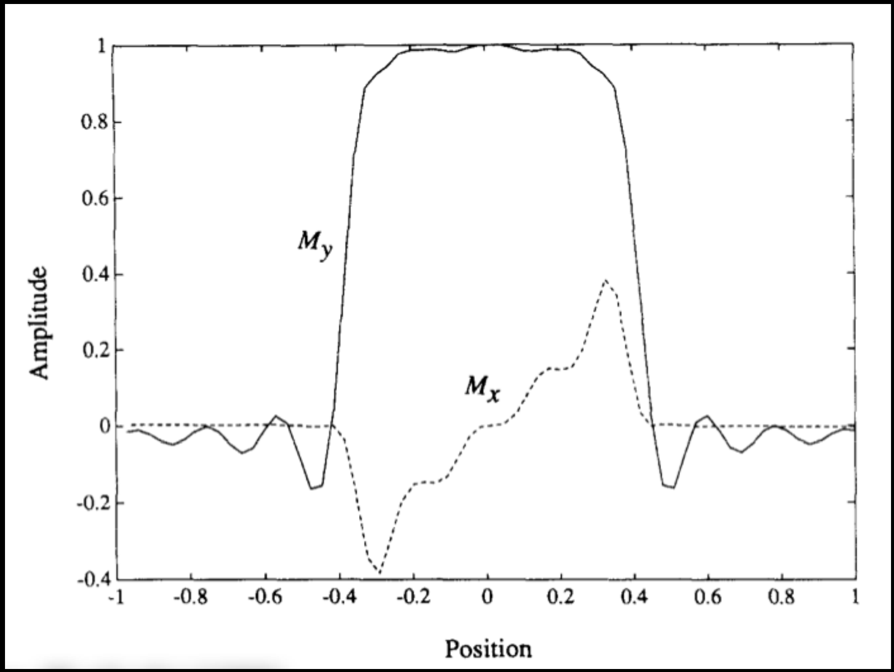
the approximation works surprisingly well even for flip angles up to 90^0

Shaped Pulses

30°



90°



Pauly, J. *J. Magn. Reson.* **81** 43-56 (1989)

small-angle approximation still works reasonably well for flip angles that aren’t necessarily “small”

Truncation Artifacts

in MRI we want pulses to be as short as possible to avoid relaxation effects

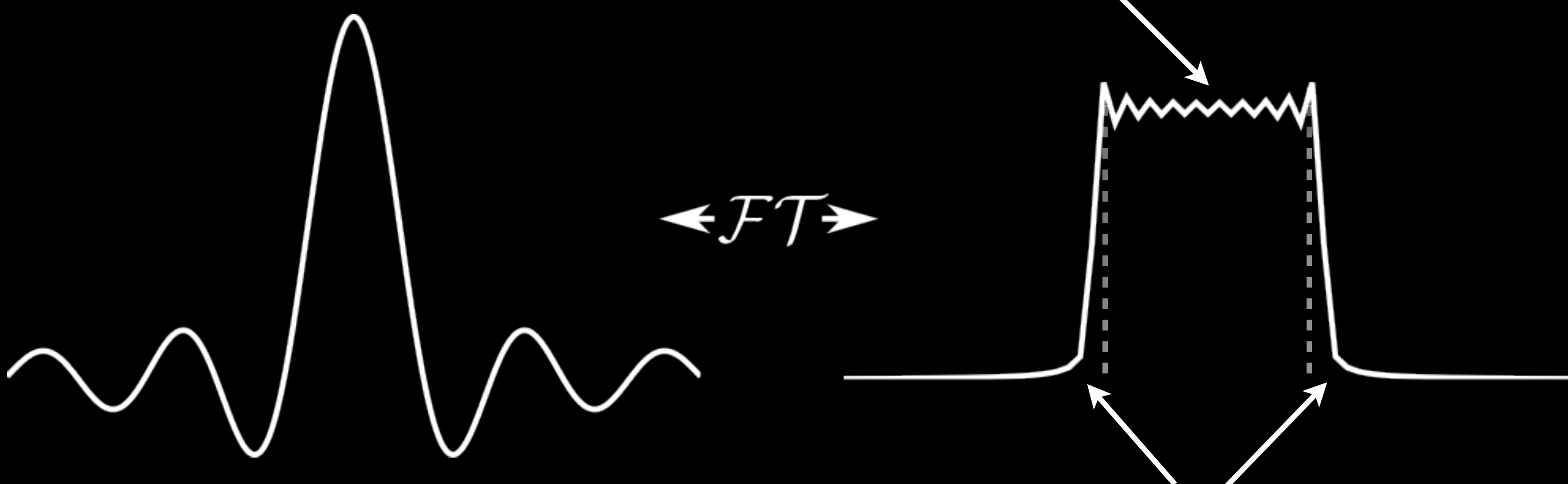
the sinc function is defined over all time which is impractical in any experiment

the sinc pulse needs to be truncated to be appropriate for clinical scans

Truncation Artifacts

what happens when we truncate our pulses?

non-uniform excitation



excited outside of slice

these deviations from the ideal are known as truncation artifacts

Truncation Artifacts

alternative Pulse Shapes

gaussian

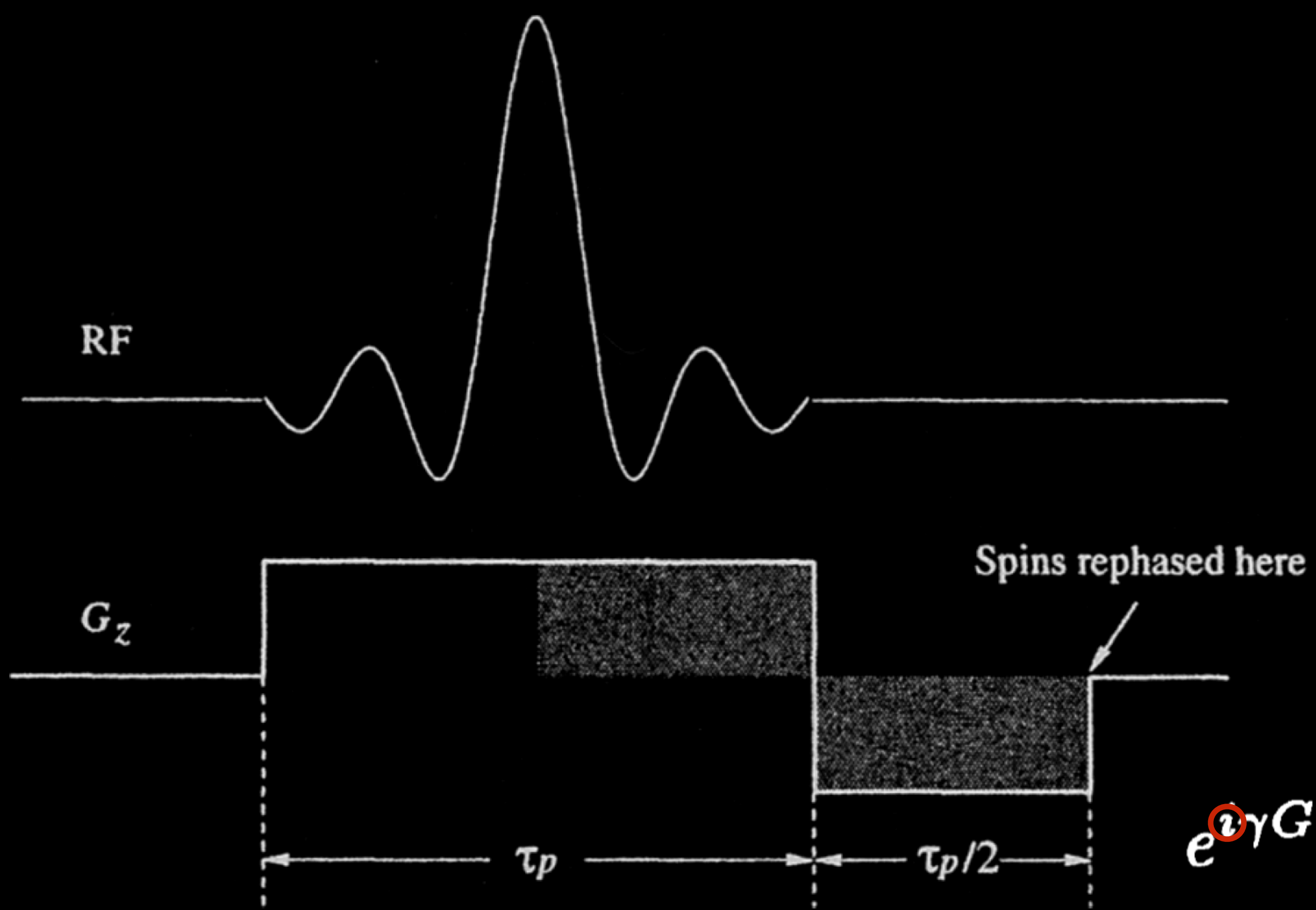
$$B_x(t) = A \exp\left[-a(t - \tau/2)^2\right]$$

reduced side-lobes, but not as flat of a profile

Window Functions

Hamming, Hanning, …

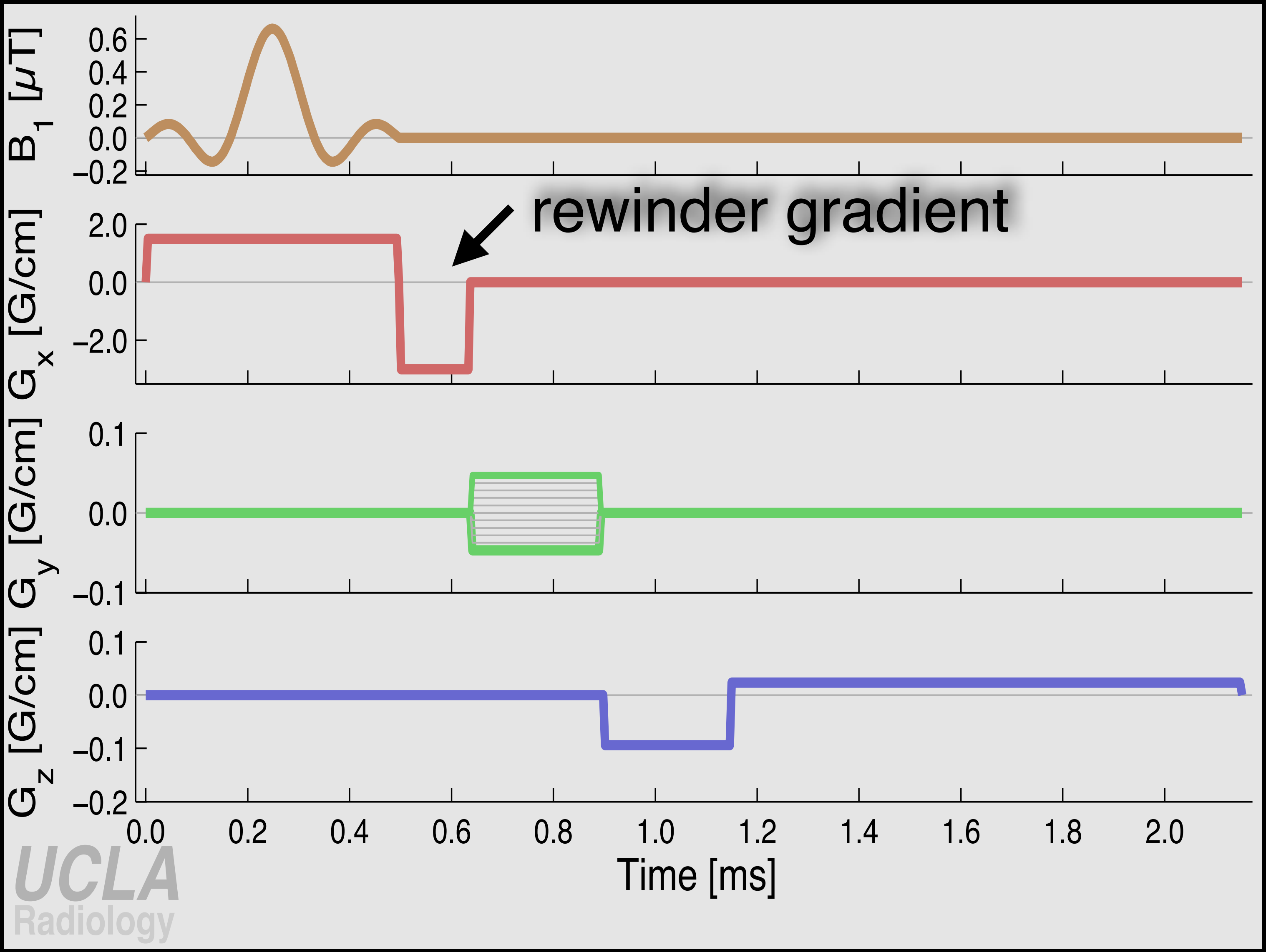
Slice Rewinder



$$e^{i\gamma G_z (z - z_0)\tau_p/2}$$

Opposite Polarity

Slice Selective Excitation Example



slice select gradient rewinder eliminates the linear phase ramp

Selective Excitation: Conclusion

B1 amplitude

-> flip angle

B1 amplitude profile

-> bandwidth, slice profile

B1 carrier frequency

-> slice location

B1 phase profile

-> slice location, etc.

Small Tip Approximation

-> slice profile = FT of B1 envelope function

MATLAB Demo

```
%% Design of Windowed Sinc RF Pulses
tbw = 4;
samples = 512;
rf = wsinc(tbw, samples);
```

```
%% Plot RF Amplitude
flip_angle = pi/2;
rf = flip_angle*rf;

pulseduration = 1;        % in msec
dt  = pulseduration/samples;
rfs = rf/(gamma*dt);      % Scaled to Gauss

bw = tbw/pulseduration; % in kHz
gmax = bw/gamma_2pi;

b1    = [rfs zeros(1,samples/2)];                    % in Gauss
g     = [ones(1,samples) -ones(1,samples/2)]*gmax;   % in G/cm
t_all = (1:length(g))*dt;  % in msec
```

MATLAB Demo

```
%% Simulate Slice Profile using Bloch Simulation
x  = (-2:.01:2);               % in cm
f  = 0;                        % in Hz
dt = pulseduration/samples/1e3;
t  = (1:length(b1))*dt;  % in usec

% Bloch Simulation
[mx,my,mz] = bloch(b1(:),g(:),t(:),1,.2,f(:),x(:),0);

% Transverse Magnetization
mxy_bloch = mx+1i*my;
```

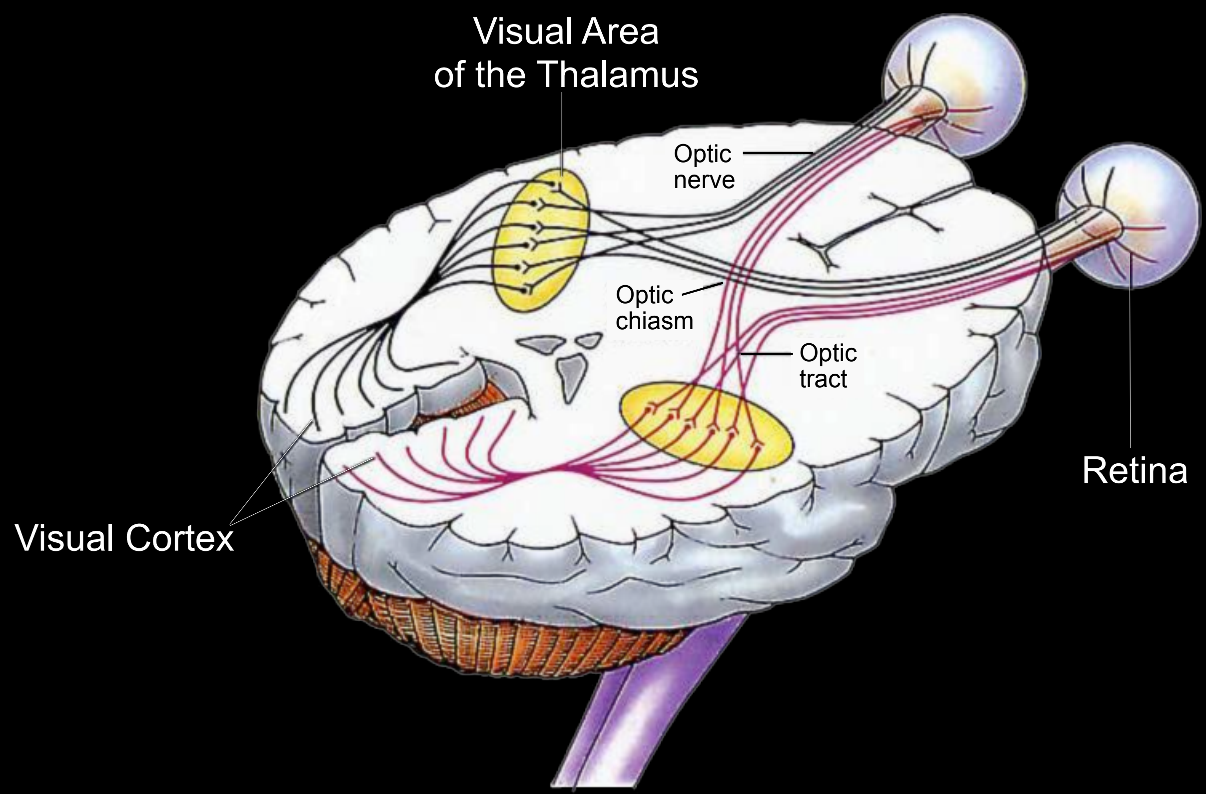
```
%% Simulate Slice Profile using Small Tip Approximation
samples_st = 4096;
f_st = linspace(-0.5/dt,0.5/dt,samples_st)/1e3;
x_st = -f_st/(gamma_2pi*gmax);

rfs_zp = zeros(1,samples_st);
rfs_zp(1:samples) = rfs;

mxy_st = fftshift(fftn(fftshift(rfs_zp)))/30;
```

Image Contrast

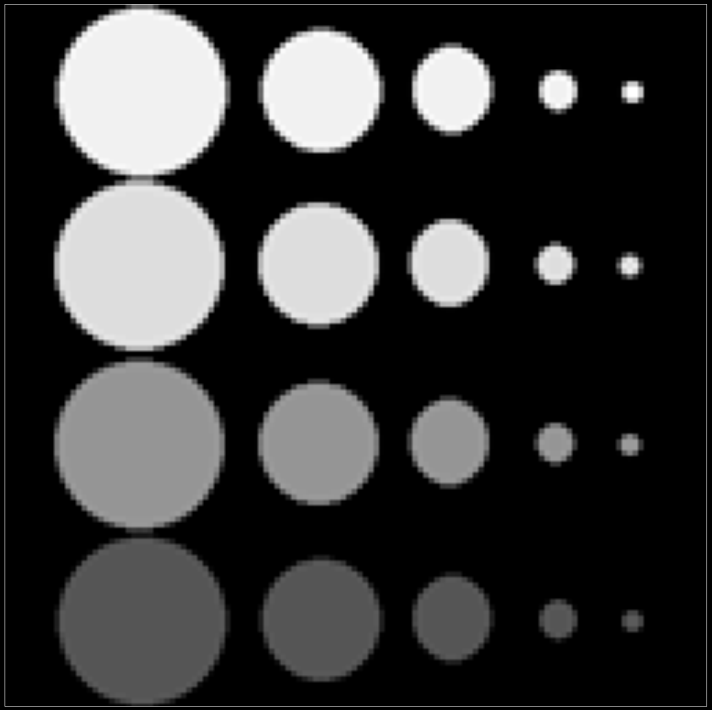
Why Image Contrast?



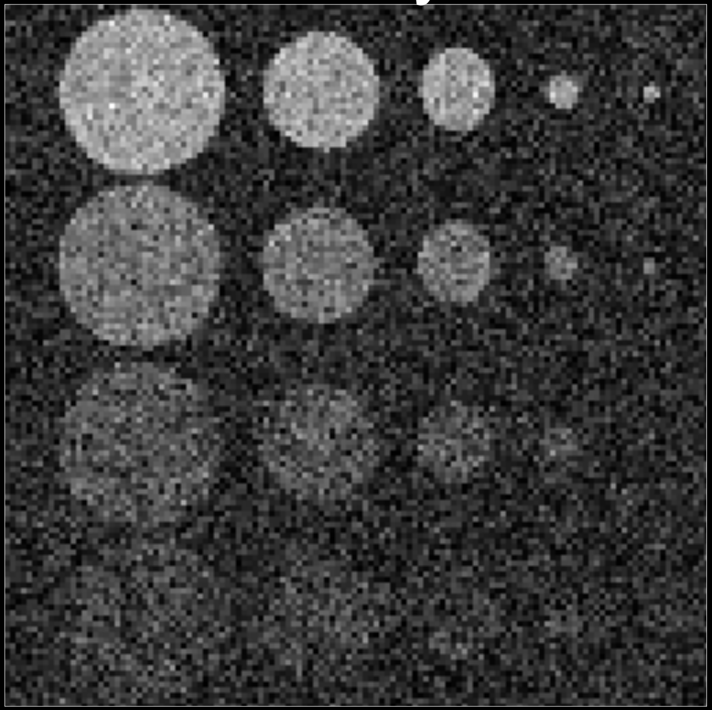
The human visual system is more sensitive to contrast than absolute luminance.

Signal to Noise Ratio (SNR)

Noise Free

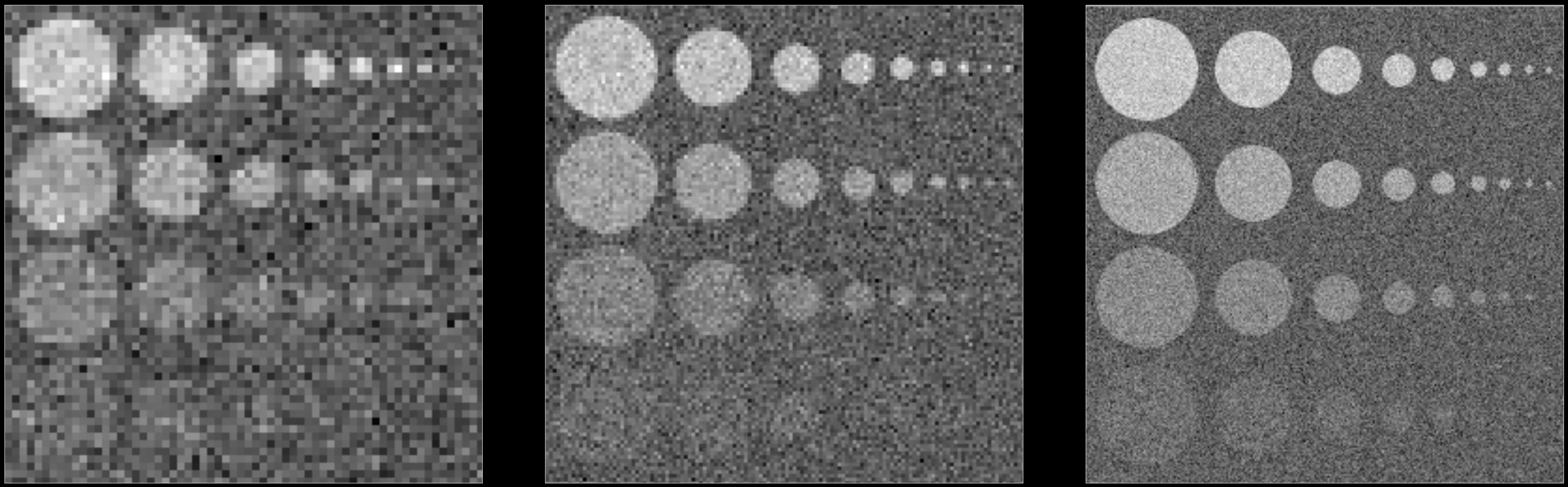


Noisy



SNR vs. Resolution

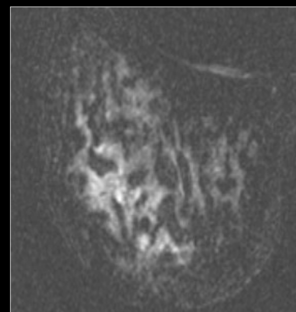
Low Resolution | Intermediate Resolution | High Resolution

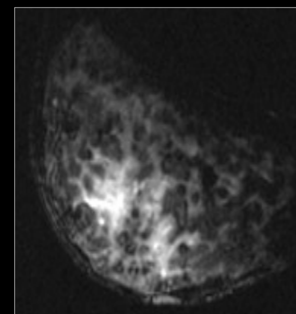


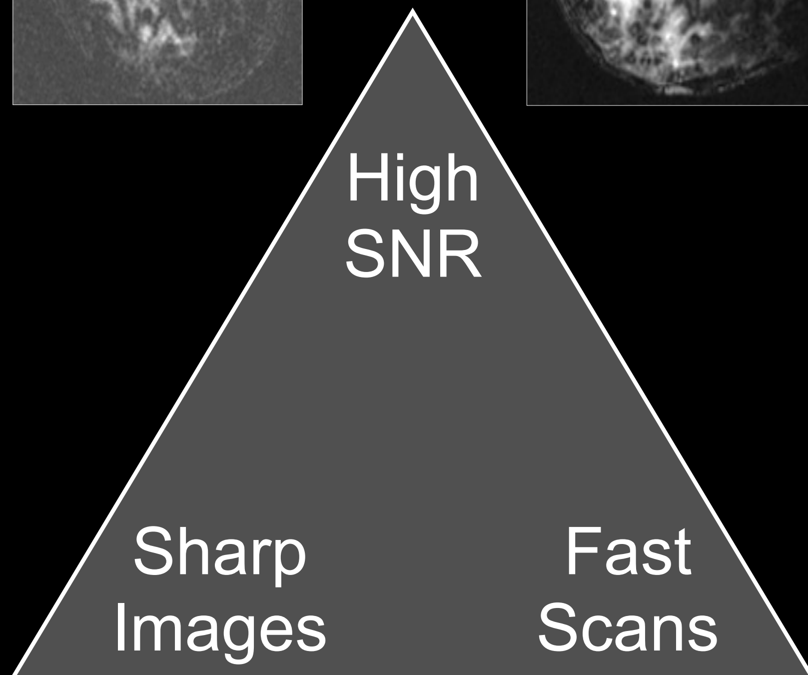
Small low-contrast objects are easier to see with higher resolution.

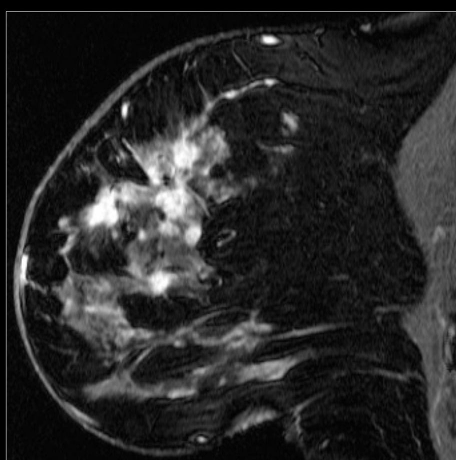
Image signal-to-noise is constant.

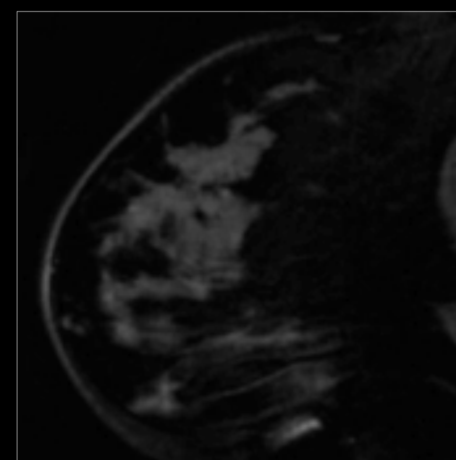
SNR vs Resolution vs Scan Time









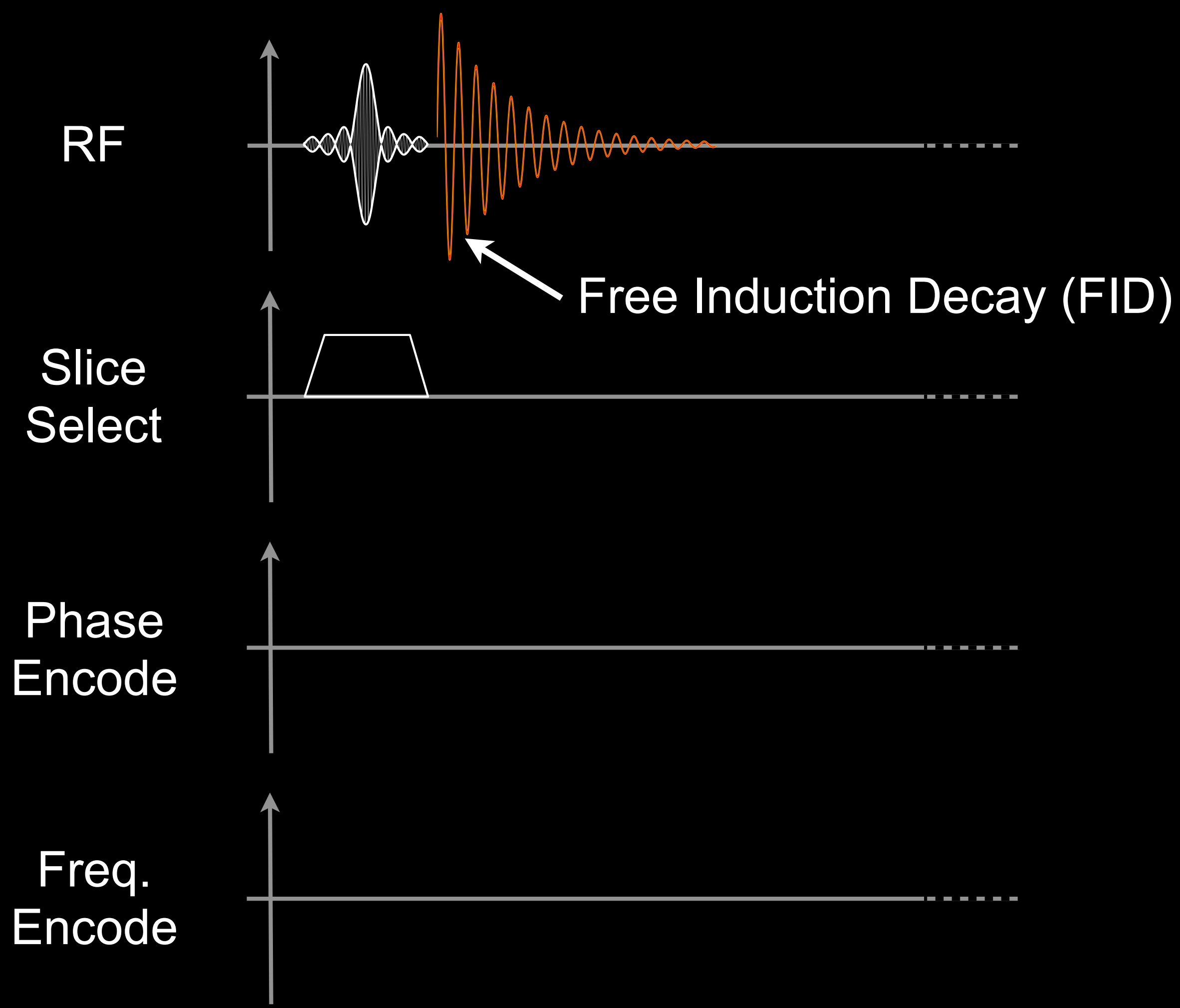


*Coils, field strength,
pulse sequence
affect starting point!*

To the Board

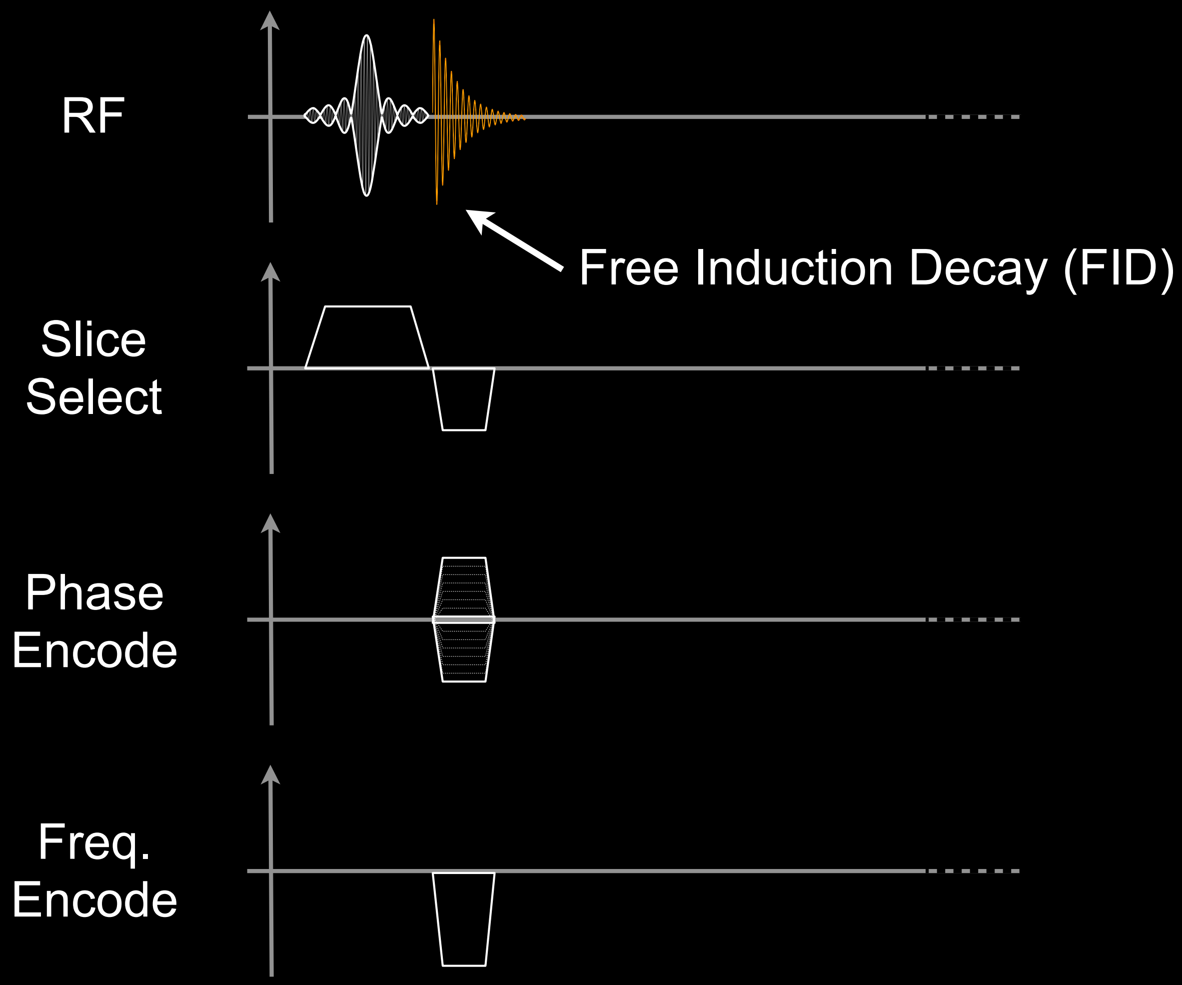
Gradient Echo Imaging

Basic Gradient Echo Sequence



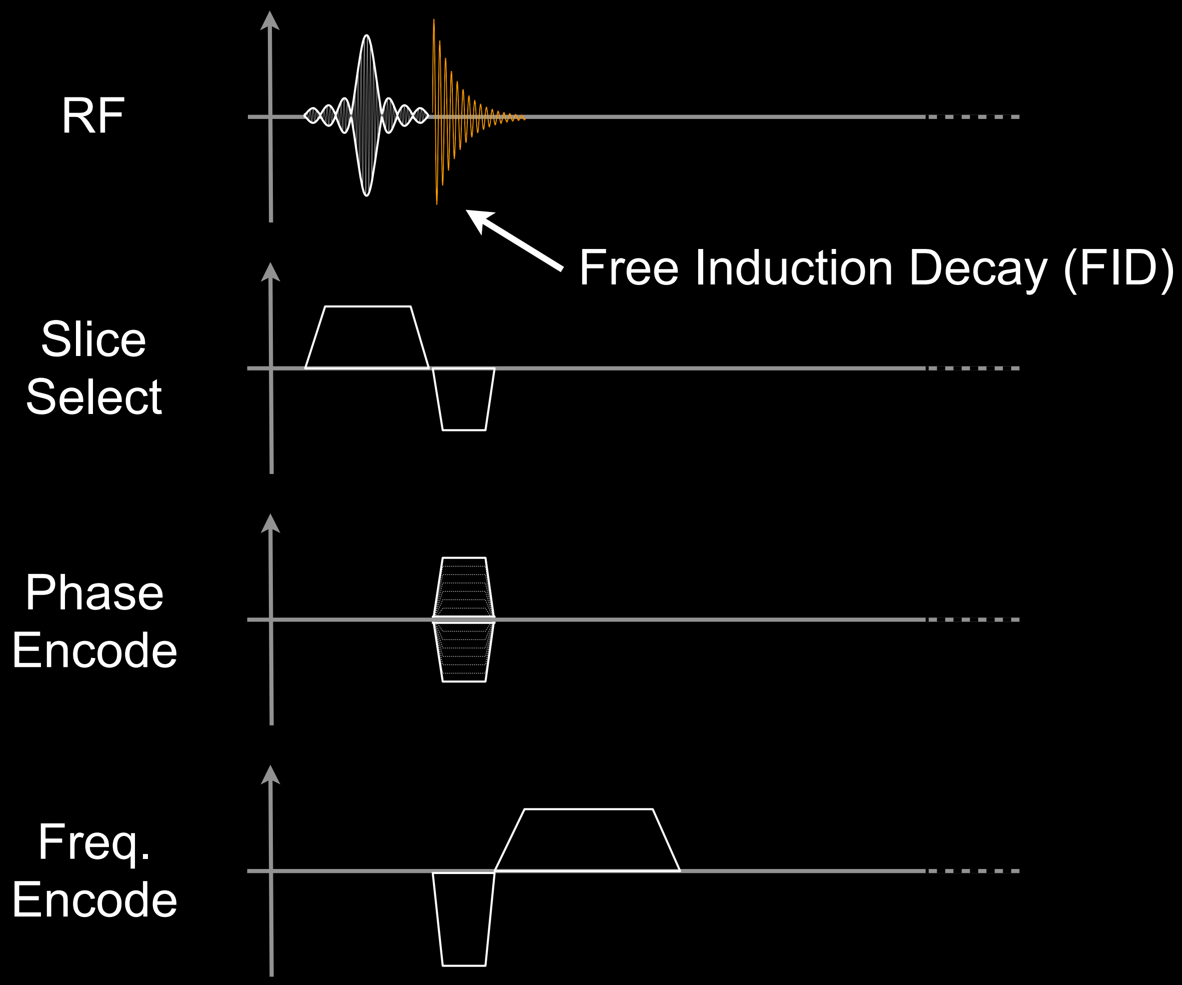
- FID Decay due to
  - T2 decay
  - Spin dephasing

Basic Gradient Echo Sequence



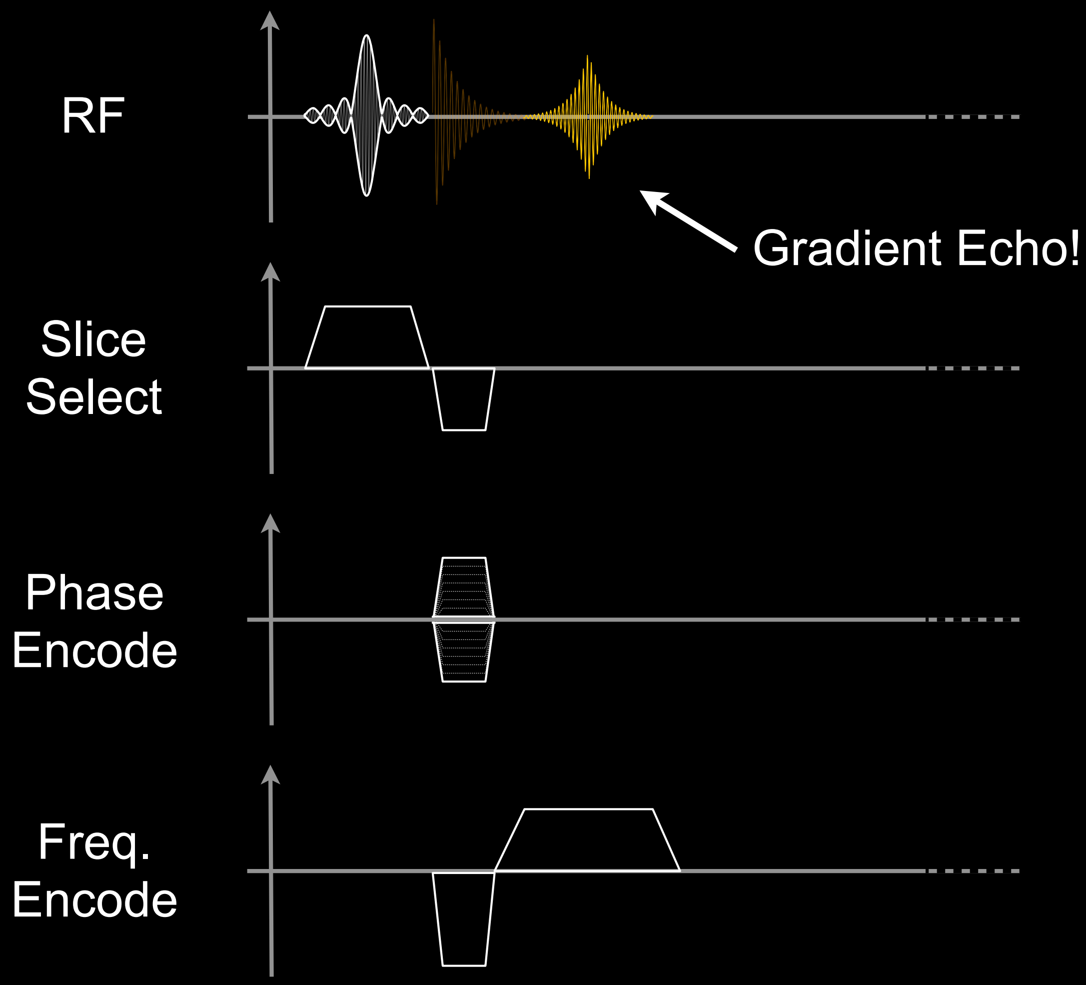
- FID Decay due to
  - T2 decay
  - Spin dephasing
- Gradients accelerate spin dephasing

Basic Gradient Echo Sequence



- FID Decay due to
  - T2 decay
  - Spin dephasing
- Gradients accelerate spin dephasing
- Gradients can undo gradient induced spin dephasing

Basic Gradient Echo Sequence



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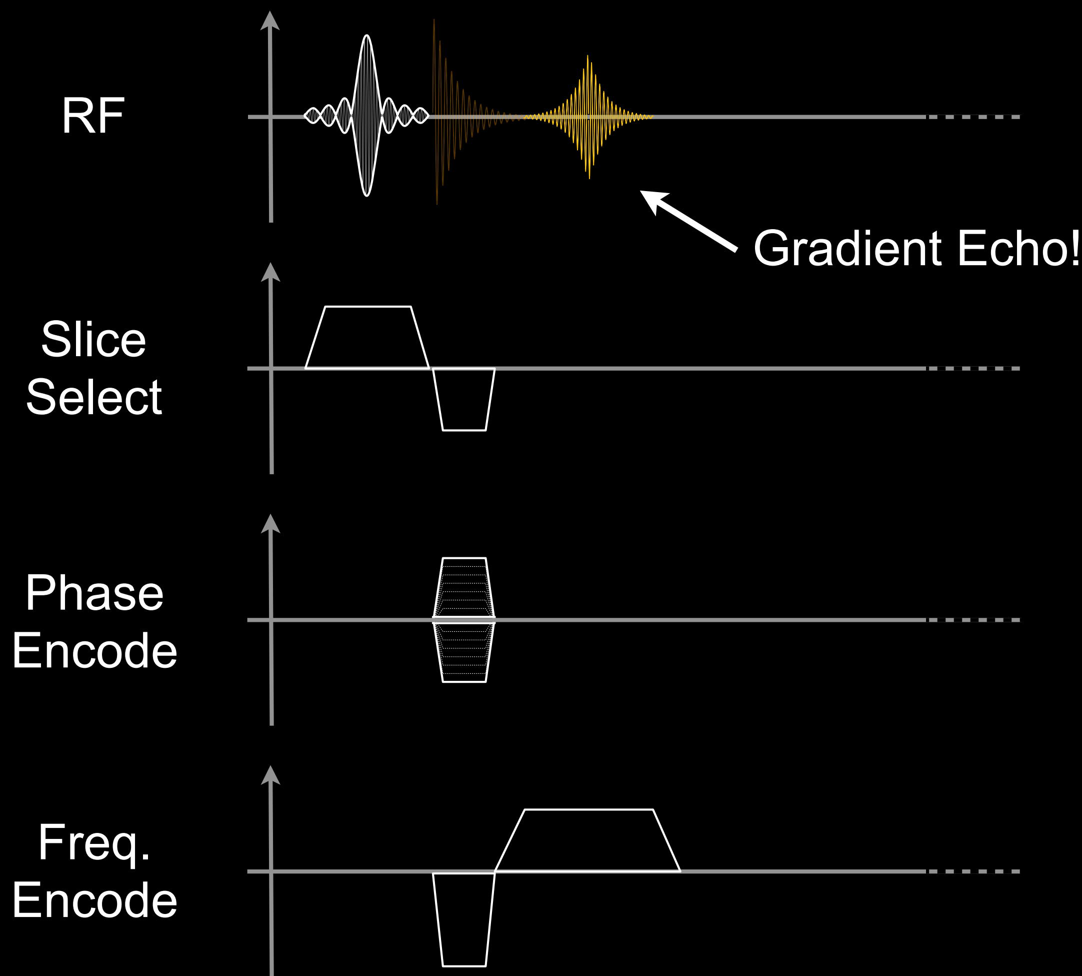

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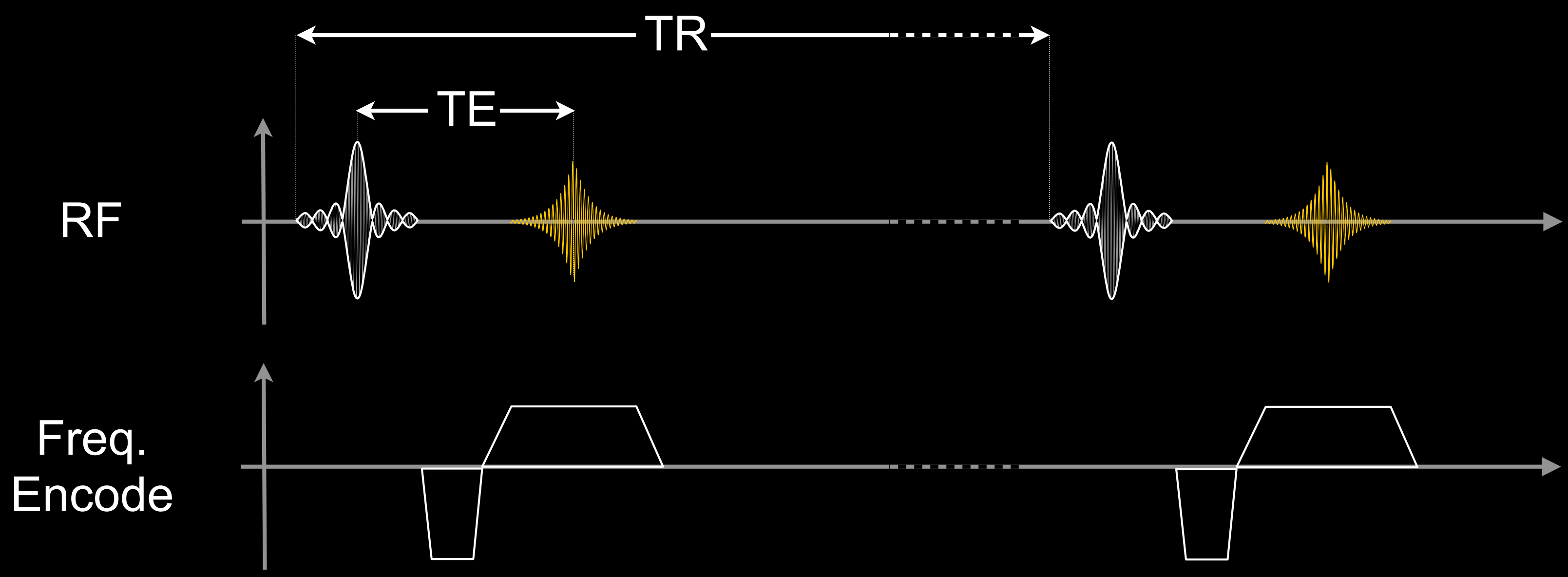
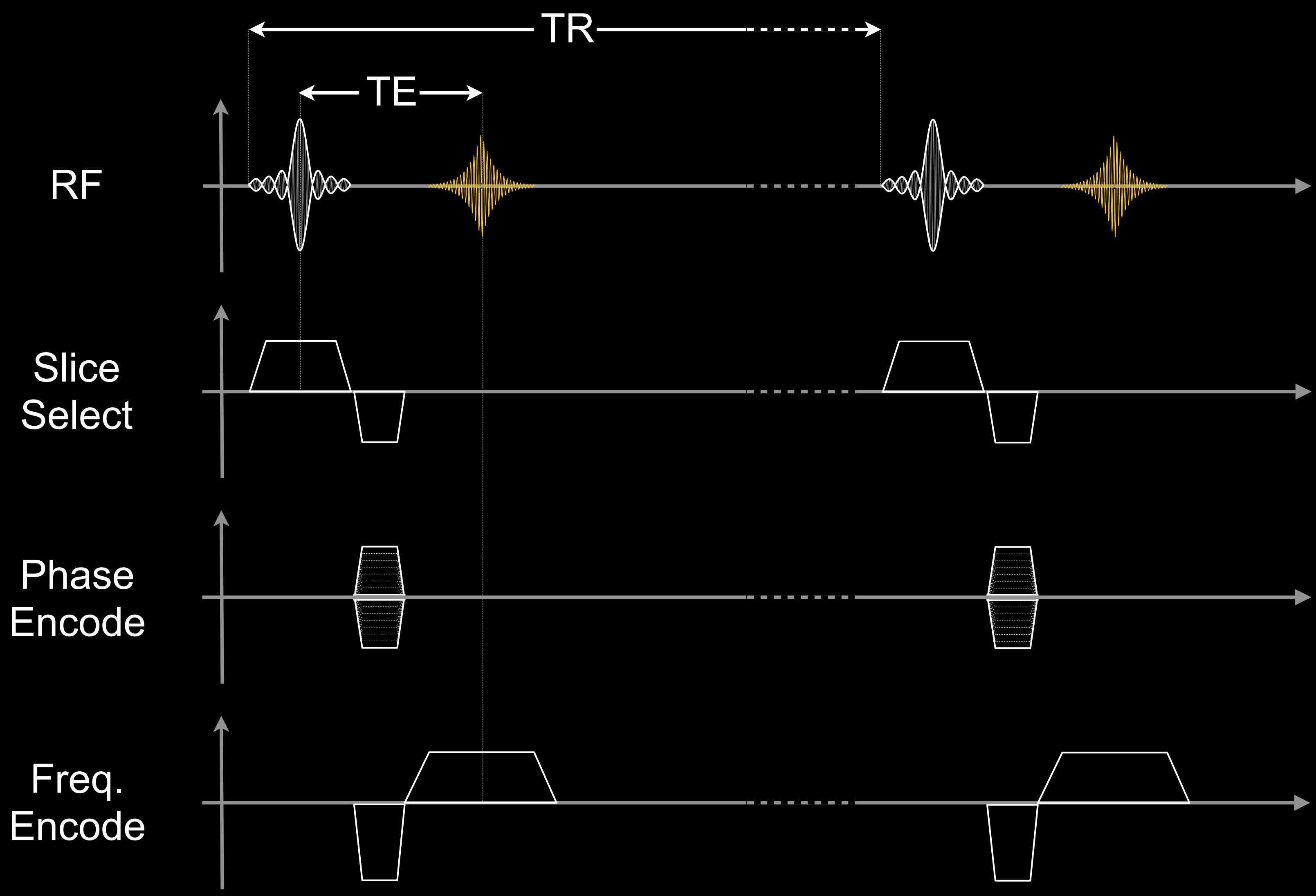
Basic Gradient Echo Sequence

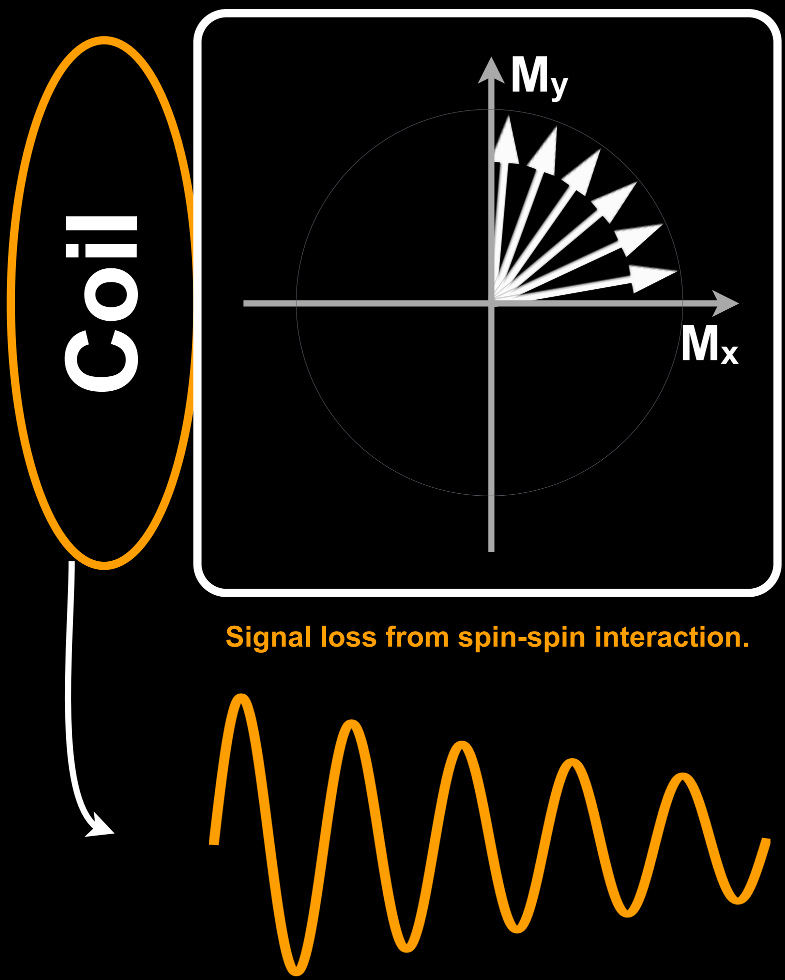


Basic Gradient Echo Sequence



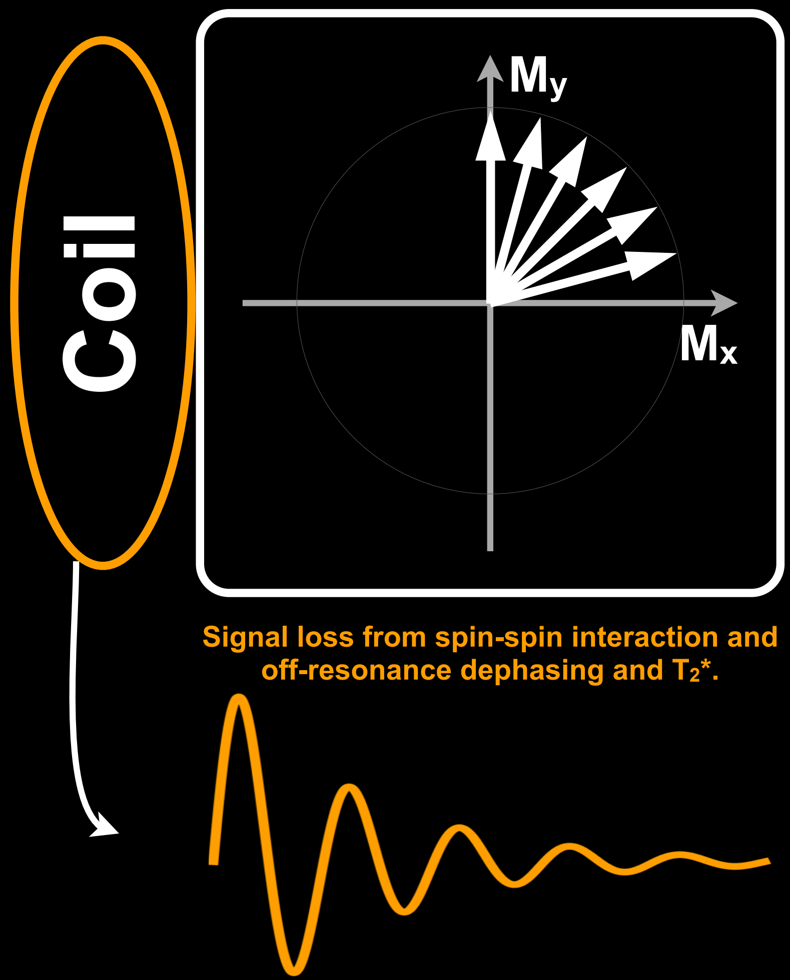
T_2 versus T_2^*

T_2 Decay



Signal loss from spin-spin interaction.

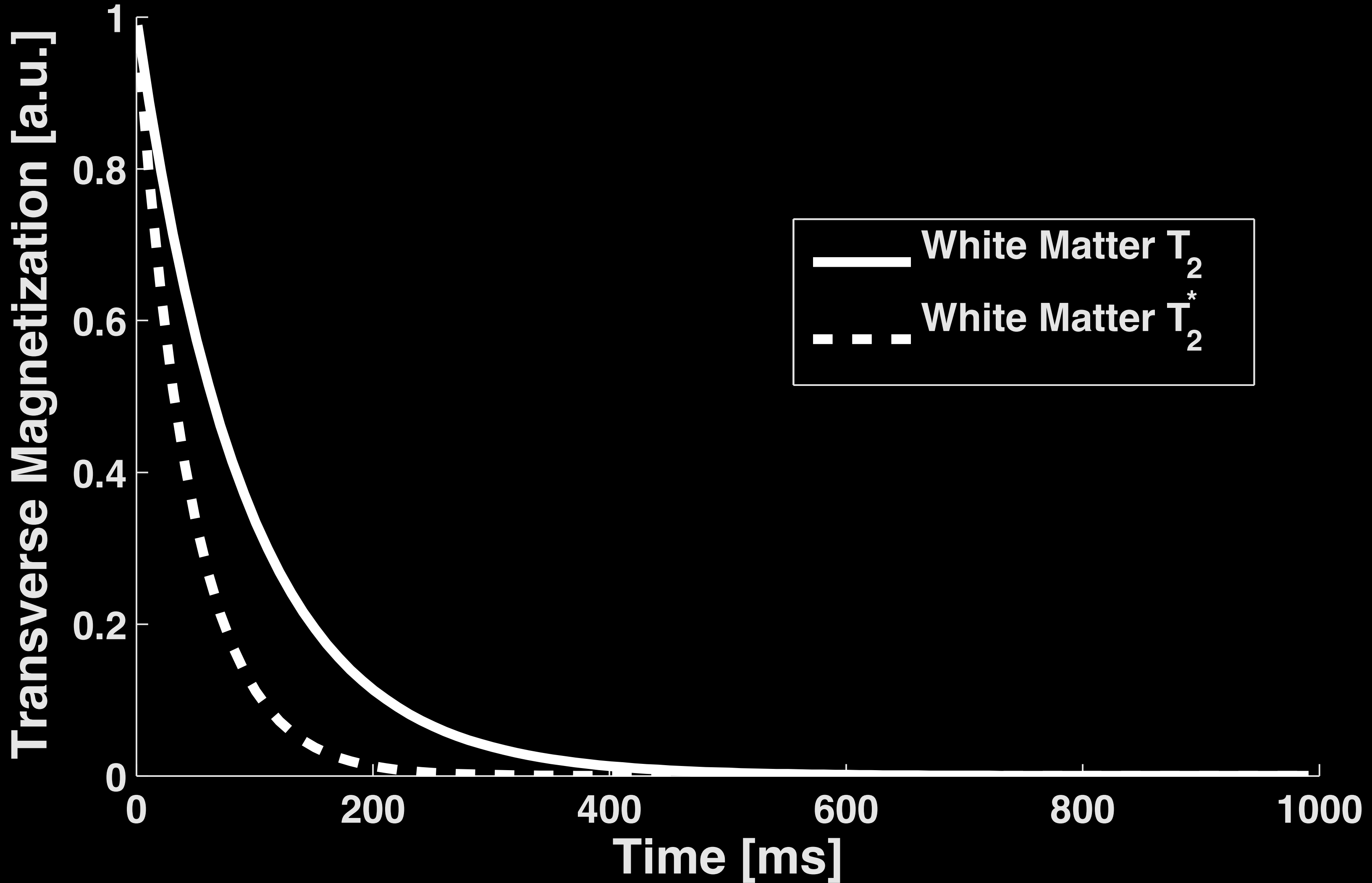
T_2^* Decay



Signal loss from spin-spin interaction and off-resonance dephasing and T_2^*.

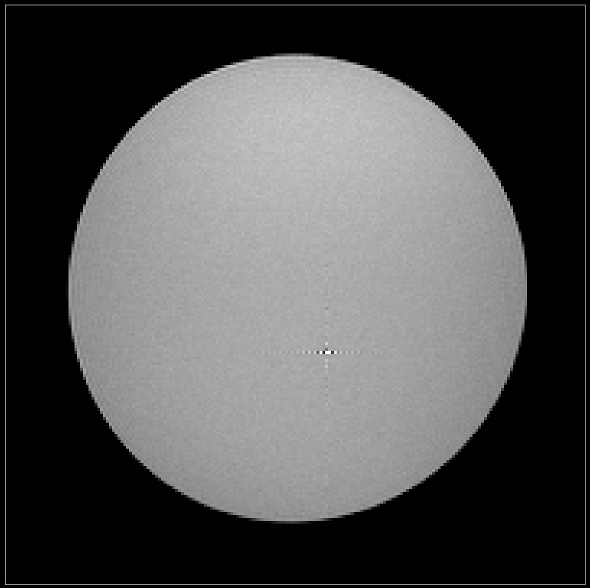
T_2^* is signal loss from spin dephasing and T_2

T2*<T2 (always!)

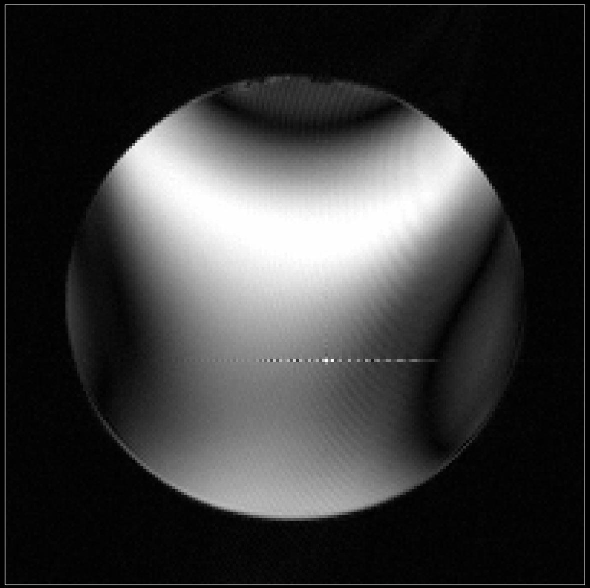

Transverse Magnetization [a.u.]
1
0.8
0.6
0.4
0.2
0
0
200
400
600
800
1000
Time [ms]
White Matter T_2
White Matter T_2^*

SE vs. GRE: B_0 Inhomogeneity

- Images acquired with a bad shim
  - Poor B_0 homogeneity (lots of off-resonance)



Spin Echo



Gradient Echo

Images Courtesy of http://chickscope.beckman.uiuc.edu/roosts/carl/artifacts.html

Gradient Echoes & Contrast

Gradient Echo Sequences

- Spoiled Gradient Echo
  - SPGR, FLASH, T1-FFE
- Balanced Steady-State Free Precession
  - TrueFISP, FIESTA, Balanced FFE

Principal GRE Advantages

- Fast Imaging Applications
  - **Why?** *Can use a shorter TE/TR than spin echo*
  - **When?** Breath-held, realtime, & 3D volume imaging
- Flexible image contrast
  - **Why?** Adjusting TE/TR/FA controls the signal
  - **When?** Characterize a tissue for diagnosis
- Bright blood signal
  - **Why?** Inflowing spins haven't "seen" numerous RF pulses
  - **When?** Cardiovascular & angiographic applications

Principal GRE Advantages

- Low SAR
  - **Why?** Imaging flip angles are (typically) small
  - **When?** When heating risks are a concern
- Quantitative
  - **Why?** Multi-echo acquisition are practical.
  - **When?** Flow quantification & Fat/Water mapping
- Susceptibility Weighted Imaging
  - **Why?** No refocusing pulse.
  - **When?** T_2^*-weighted (hemorrhage) imaging
- More...

Principal GRE Disadvantages

- Off-resonance sensitivity
  - **Why?** No refocusing pulse
    - Field inhomogeneity, Susceptibility, & Chemical shift

- T_2^*-weighted rather than T_2-weighted
  - **Why?** No re-focusing pulse
    - Spin-spin dephasing is not reversible with GRE

- Larger metal artifacts than SE
  - Why? No refocusing pulse.
    - Large field inhomogeneities aren't corrected with GRE

Spoiled Gradient Echo Contrast

Contrast depends on tissue's ρ, T_1 and T_2*.

$$A_{echo} \propto \frac{\rho\left(1 - e^{-TR/T_1}\right)}{1 - \cos\alpha e^{-TR/T_1}} \sin\alpha e^{-TE/T_2^*}$$

Contrast adjusted by changing TR, flip angle, and TE

Spoiled Gradient Echo Contrast

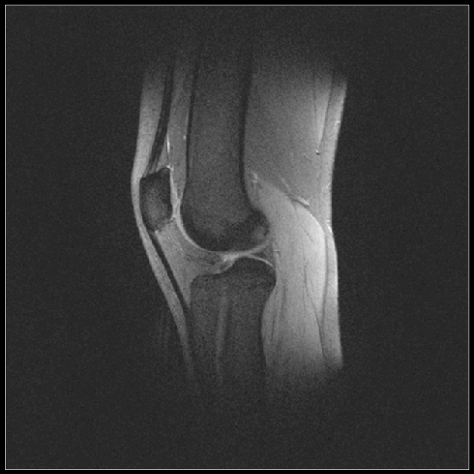
Gradient Echo Parameters

Type of Contrast	TE	TR	Flip Angle
Spin Density	Short	Long	Small
T_1-Weighted	Short	Intermediate	Large
T_2^*-Weighted	Intermediate	Long	Small

T_2*-weighted Gradient Echo MRI

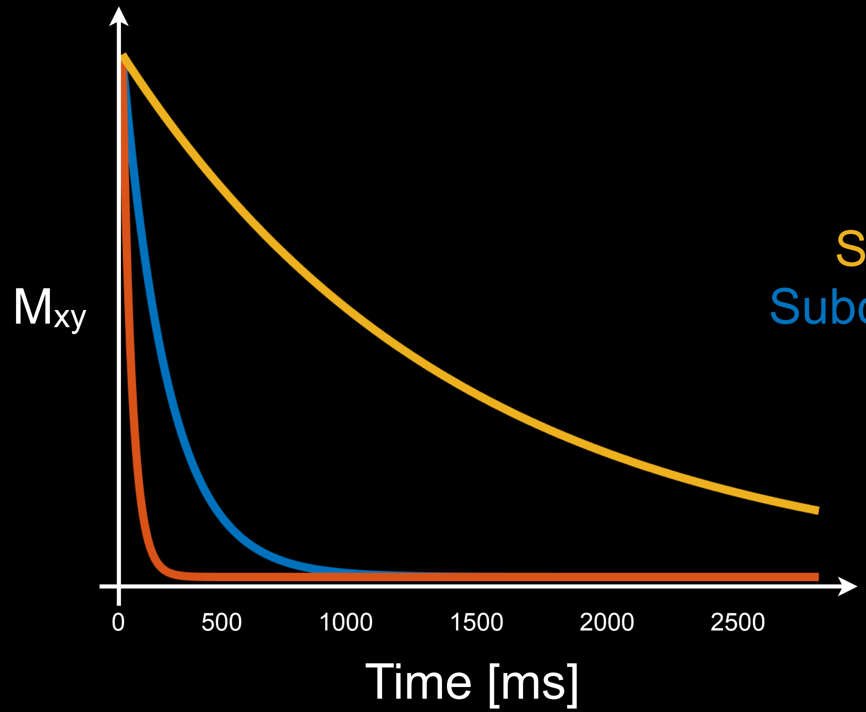
FLASH – TE=4.8ms; TR=200ms

FLASH – TE=14.2ms; TR=200ms



FLASH – TE=24ms; TR=200ms

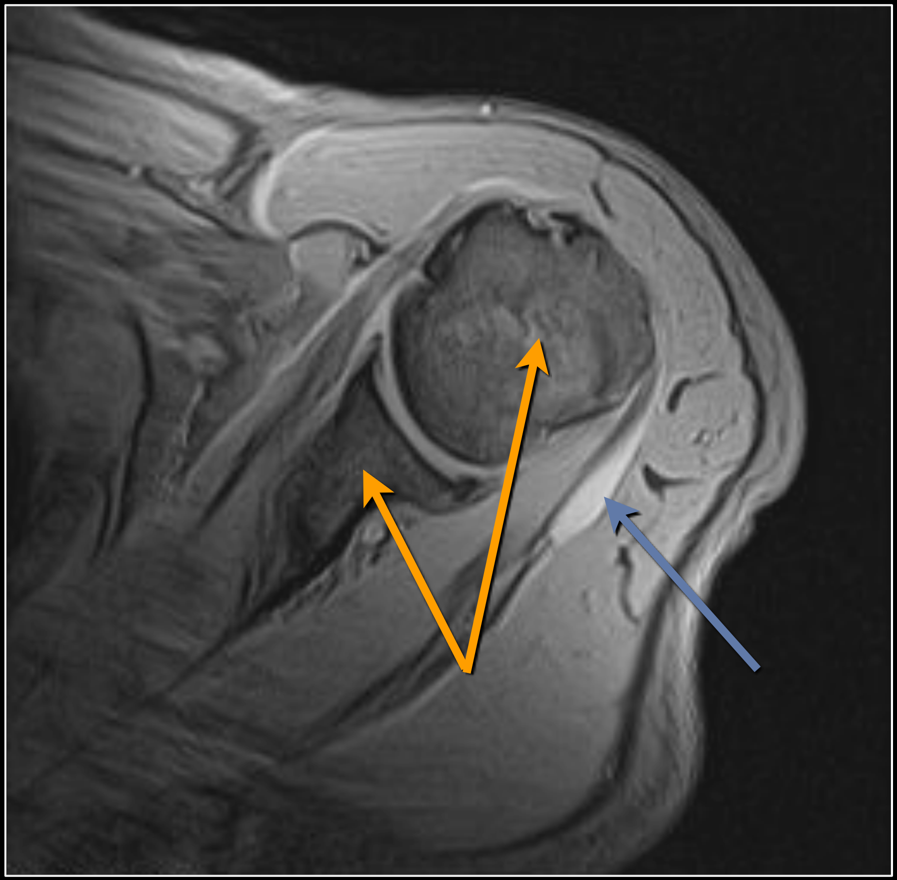
FLASH – TE=49ms; TR=200ms



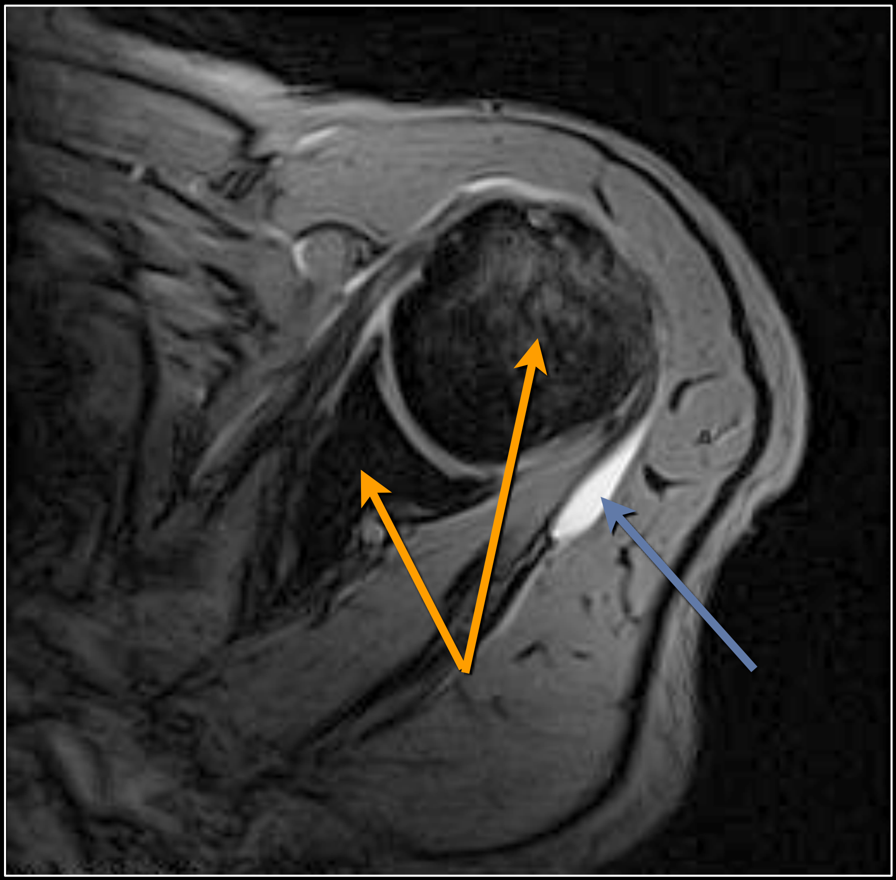
Synovial Fluid T_2~1210ms
Subcutaneous Fat T_2~165ms
Muscle T_2~35ms

Musculoskeletal MRI at 3.0 T: relaxation times and image contrast. AJR Am J Roentgenol. 2004 Aug;183(2):343-51.

T_2*-weighted Gradient Echo MRI



TE=9ms



TE=30ms

Susceptibility Weighting (darker with longer TE)

Bright fluid signal (long T_2* is "brighter" with longer TE)

Images Courtesy of Brian Hargreaves

Gradient vs Spin Echo Contrast

Gradient Echo Parameters

Type of Contrast	TE	TR	Flip Angle
Spin Density	<5ms	>100ms	<10°
T_1-Weighted	<5ms	<50ms	>30°
T_2^*-Weighted	>20ms	>100ms	<10°

Spin Echo Parameters

Type of Contrast	TE	TR	Flip Angle
Spin Density	10-30ms	>2000ms	90+180
T_1-Weighted	10-30ms	450-850ms	90+180
T_2-Weighted	>60ms	>2000ms	90+180

Questions?

- Related reading materials
  - Nishimura - Chap 6 and 7

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