Introduction to MRI

Part 1: Image Formation
Part 2: Image Contrast

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David Geffen School of Medicine

UCLA Radiology

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Part 1: Image Formation
Part 2: Image Contrast

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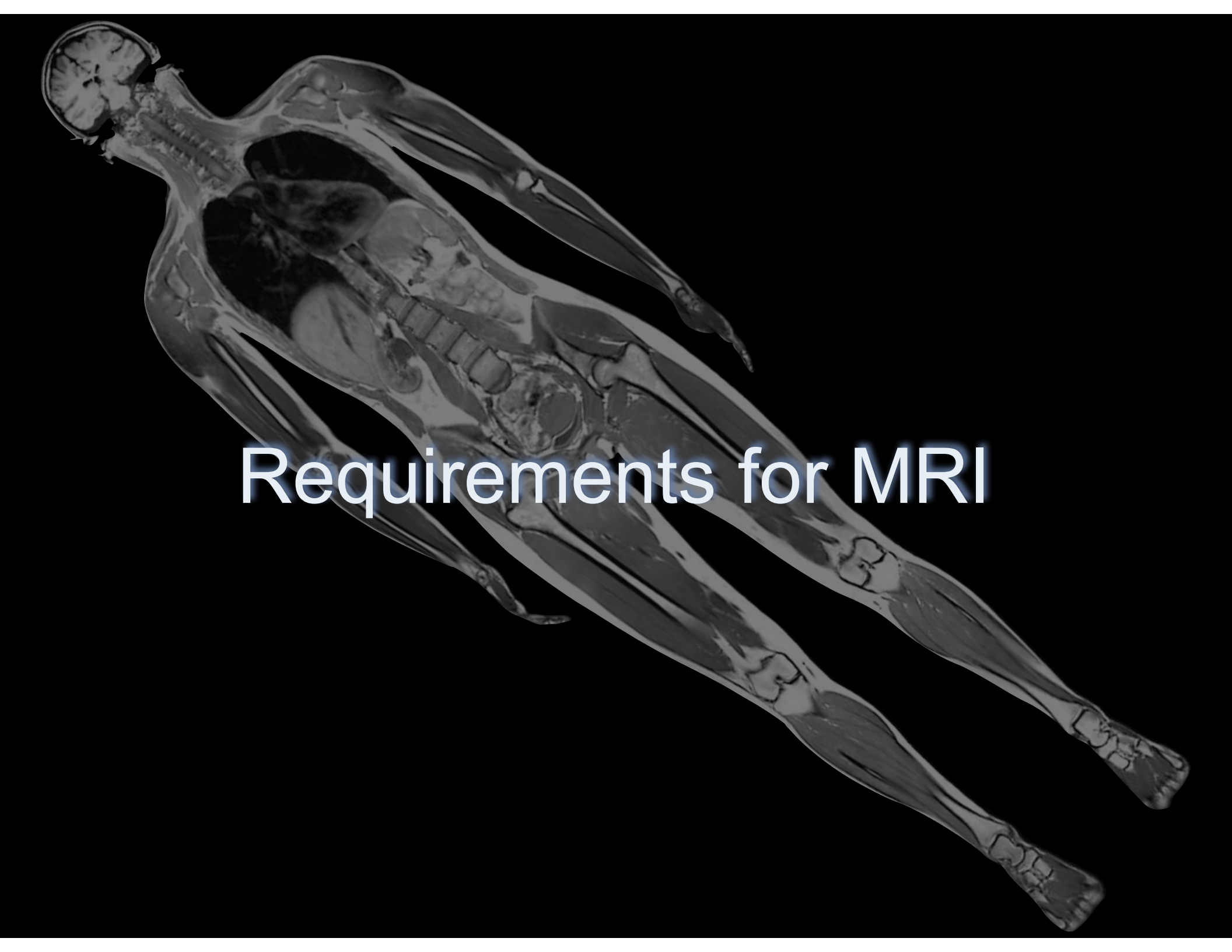

Requirements for MRI

Requirements for MRI

- NMR Active Nuclei
  - e.g. ^{1}H in H_20
- Cryogen
  - Liquid He and N_2
- Magnetic Field (B_0)
  - Polarizer
- RF System (B_1)
  - Exciter
- Coil
  - Receiver
- Gradients (G_X, G_Y, G_Z)
  - Spatial Encoding

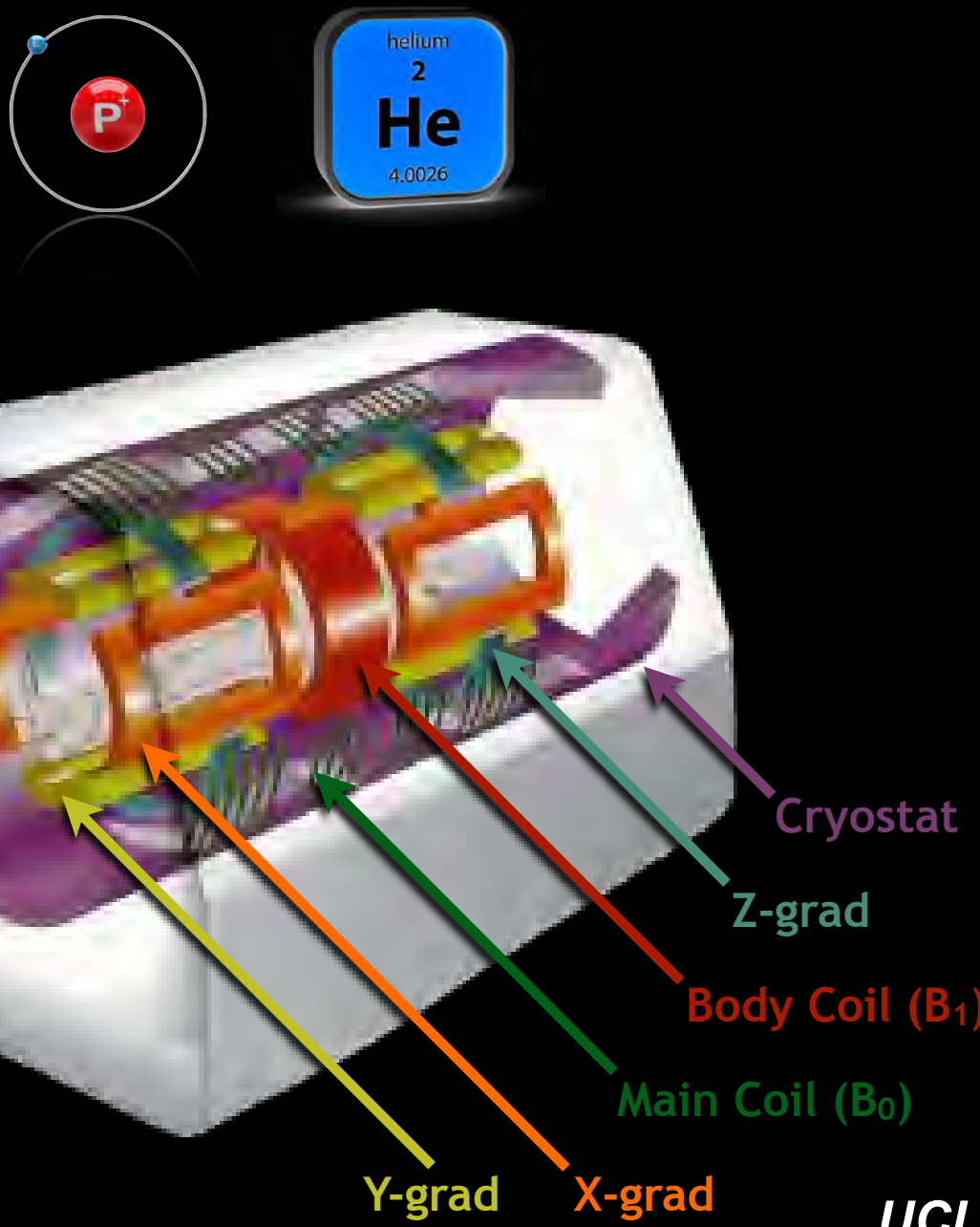
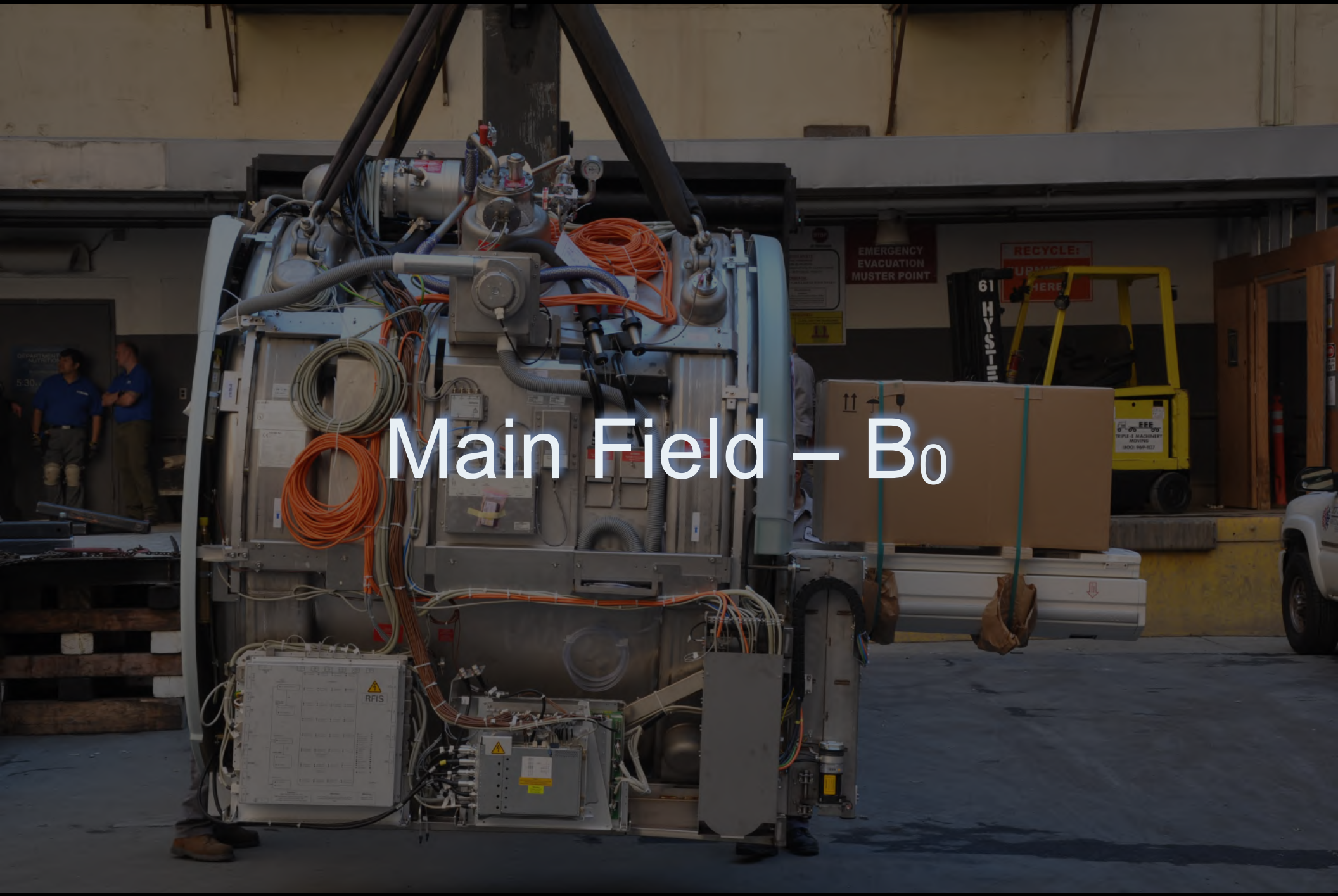


Image Adapted From: http://www.ee.duke.edu/~jshorey

Dipoles to Images

$\vec{\mu}$	Magnetic Moment	B_0
$\downarrow$		
$\vec{M}_z$	Bulk Magnetization	B_1
$\downarrow$		
$\vec{M}_{xy}$	Transverse Magnetization	Coil
$\downarrow$		
$S(t)$	Received Signal	Gradients
$\downarrow$		
$S\left(\vec{k}\right)$	k-space signal	FFT
$\downarrow$		
$I(\vec{x})$	Image	



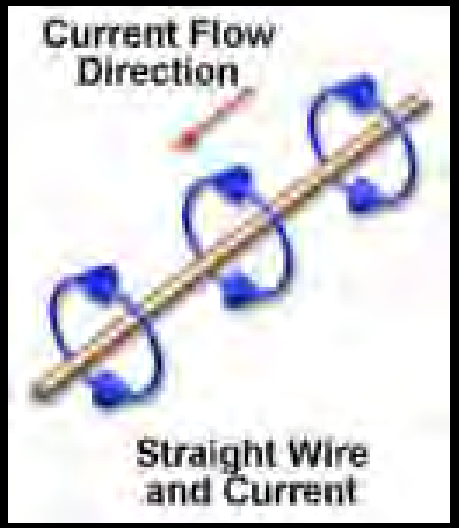
Main Field – B_0

Dipoles to Images

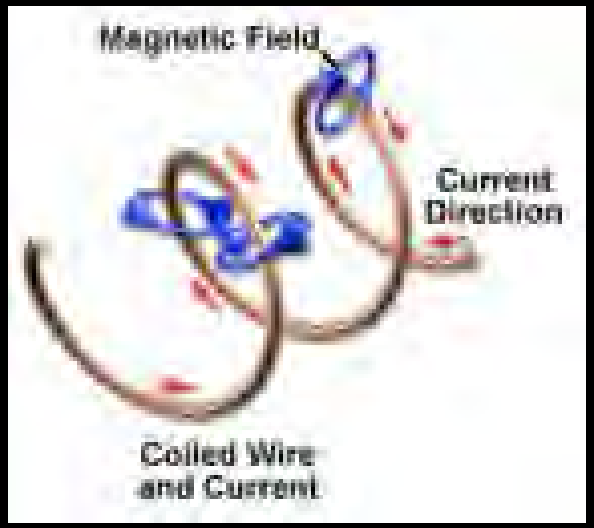
$\vec{\mu}$	Magnetic Moment	B_0
$\downarrow$		
$\vec{M}_z$	Bulk Magnetization	
$\downarrow$		B_1
$\vec{M}_{xy}$	Transverse Magnetization	
$\downarrow$		Coil
$S(t)$	Received Signal	
$\downarrow$		Gradients
$S\left(\vec{k}\right)$	k-space signal	
$\downarrow$		FFT
$I(\vec{x})$	Image	

What is B_0?

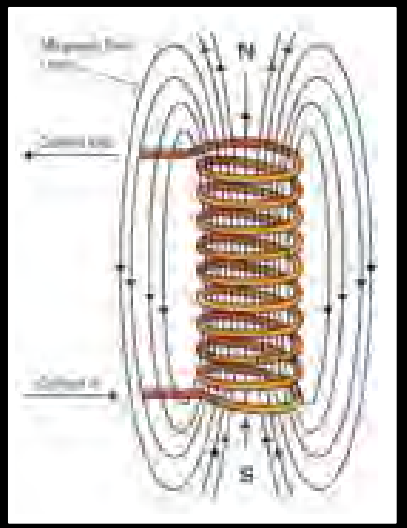
- **Superconducting Electromagnet**



Righthand Rule



Form a solenoid…



$$B = \mu I N L^{-1}$$

Superconducting Wire

- **I – Current [A]**
- **N – Number of turns**
- **L – Solenoid length [m]**
- **μ – Permeability [T•m•A^{-1}]**
- ***Example*:**
  - **1.5T=4π×10^{-7}•477.6 A•250•1 m^{-1}**

Main Field (B_0) - Principles

- B_0 is a strong magnetic field
  - 1.5T, 3.0T, 7.0T, etc.
  - Z-oriented

$$\vec{B}_0 = B_0\vec{k}$$

- B_0 forces $\vec{M}$ to precess
  - Larmor Equation

$$\omega = \gamma B$$

- B_0 generates $\vec{M}$
  - More B_0, more $\vec{M}$

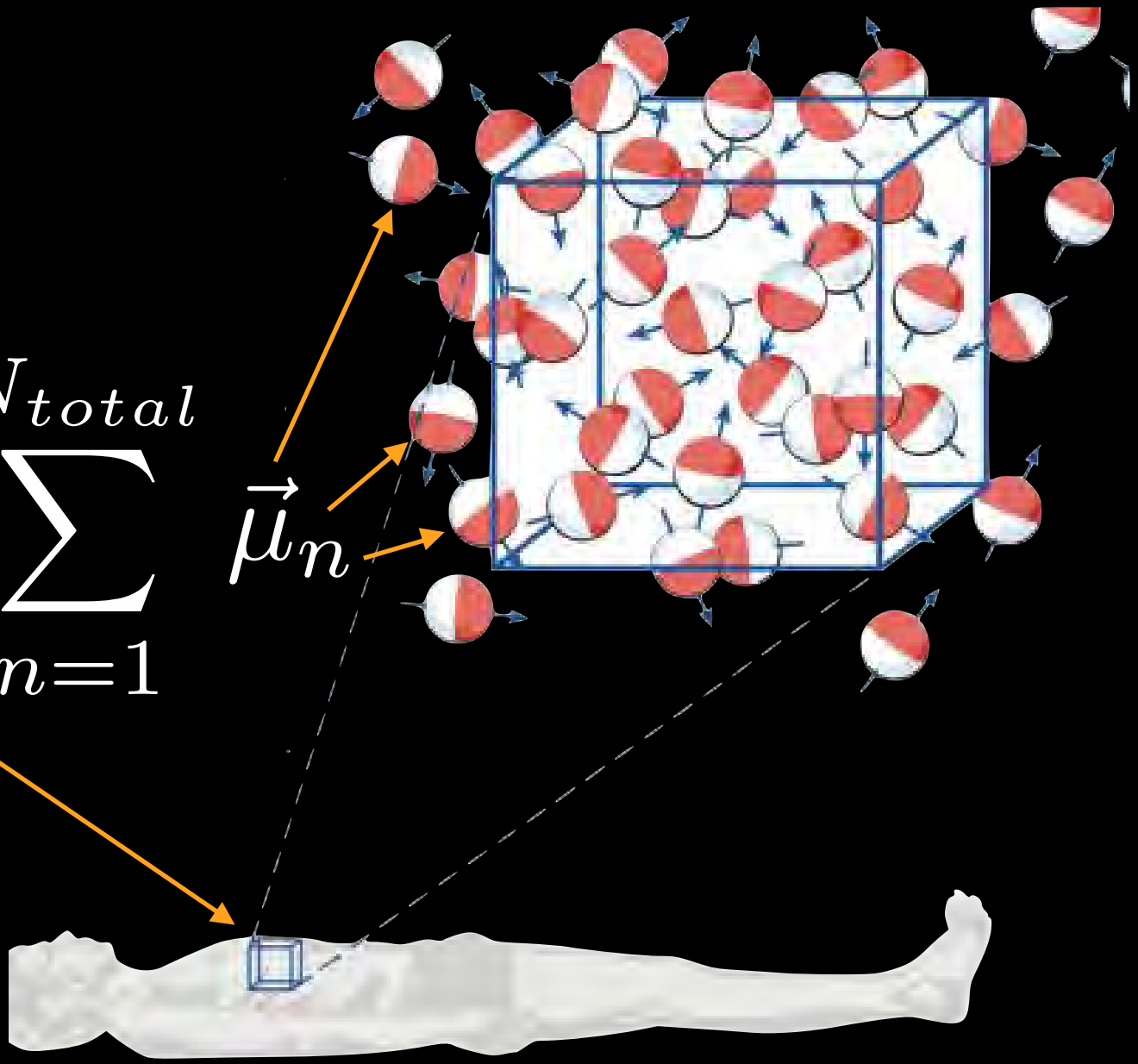
$$\vec{M} = \sum_{n=1}^{N_{total}} \vec{\mu}_n$$

Magnetic Dipoles & Larmor

Movie from Don Plewes

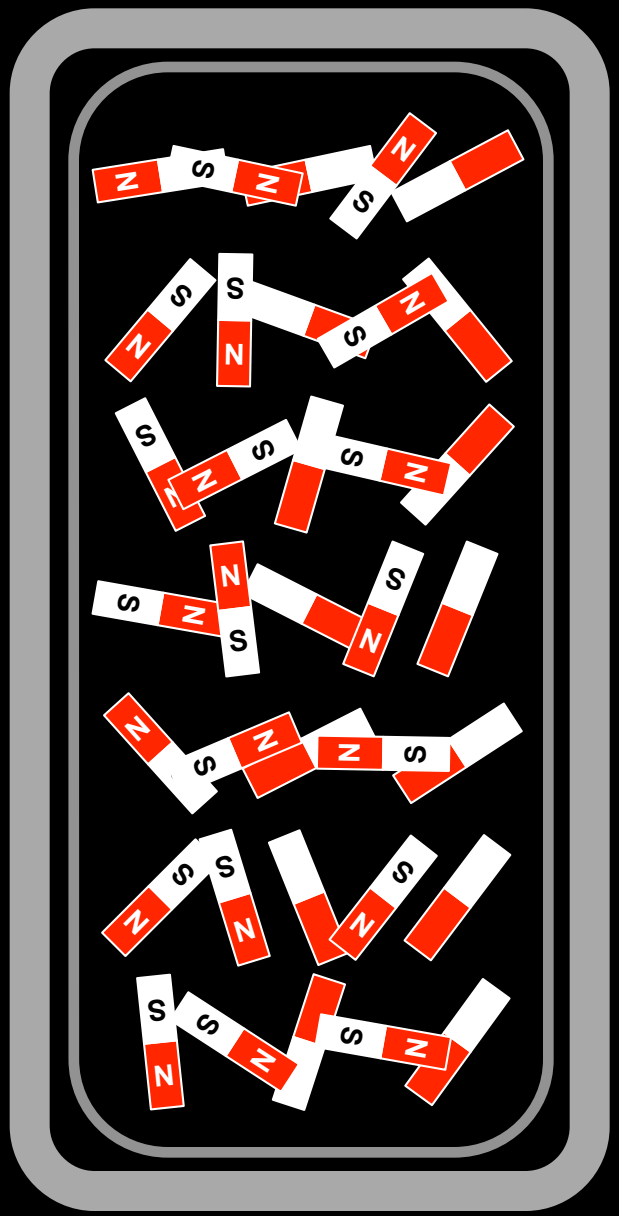
Bulk Magnetization

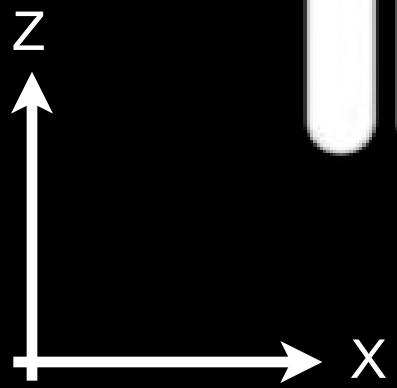
$$\vec{M} = \sum_{n=1}^{N_{total}} \vec{\mu}_n$$



N_{total}=0.24x10^{23} spins in a 2x2x10mm voxel

B_0 Field OFF





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Spins point in all directions.

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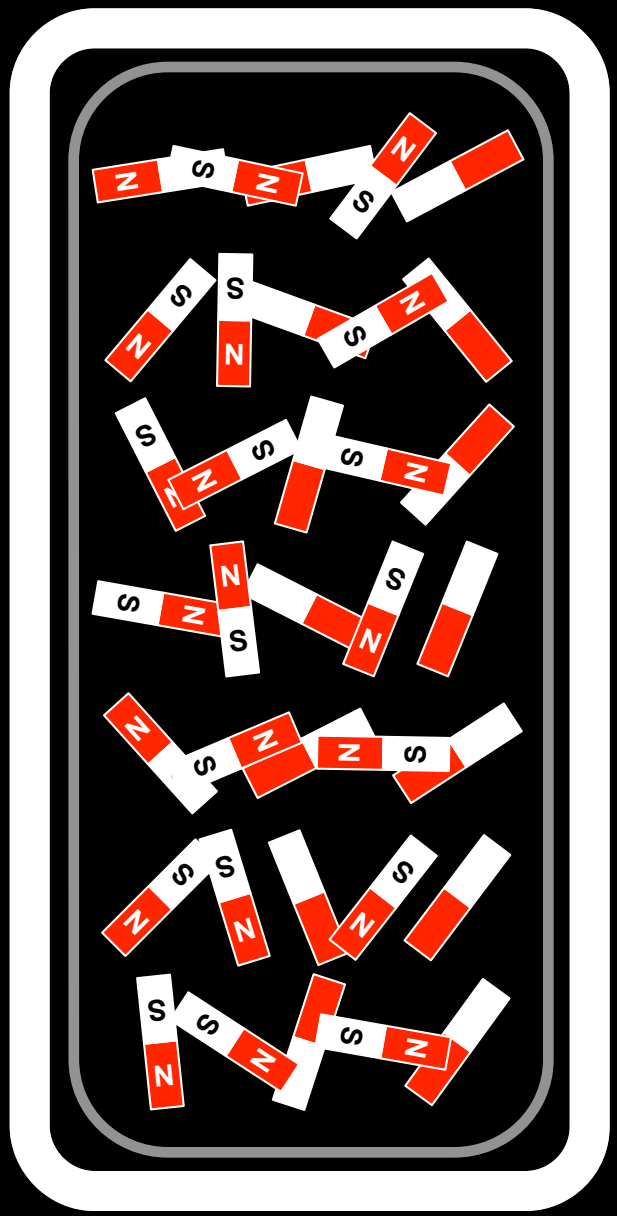
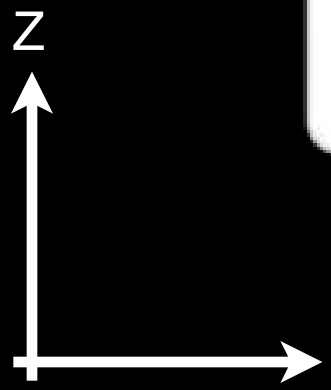
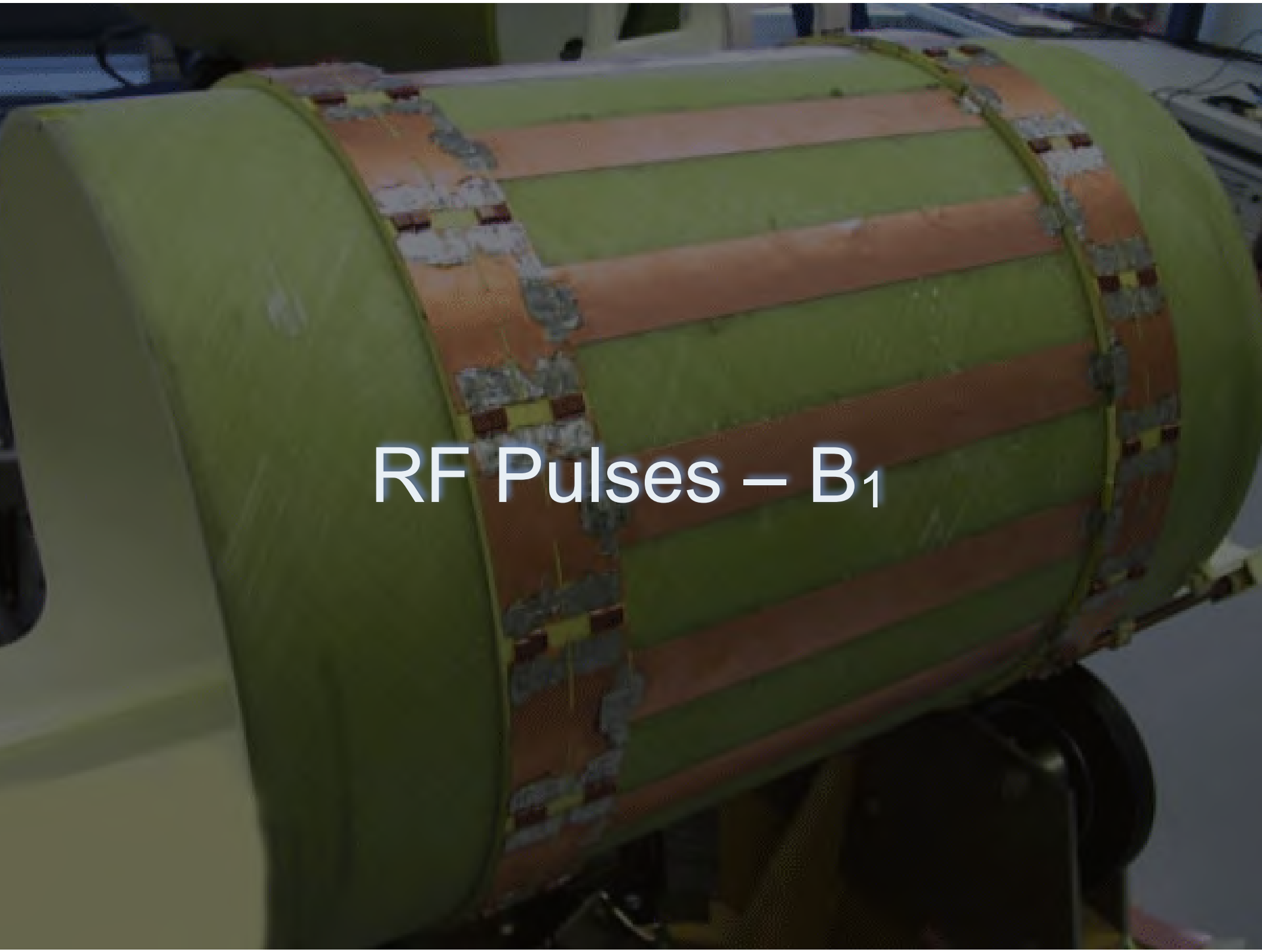
B_0 Field ON






RF Pulses – B_1

Dipoles to Images

$\vec{\mu}$ Magnetic Moment

$\downarrow$

$\vec{M}_z$ Bulk Magnetization

$\downarrow$

$\vec{M}_{xy}$ Transverse Magnetization

$\downarrow$

$S(t)$ Received Signal

$\downarrow$

$S(\vec{k})$

$\downarrow$

$I(\vec{x})$ Image

B_0

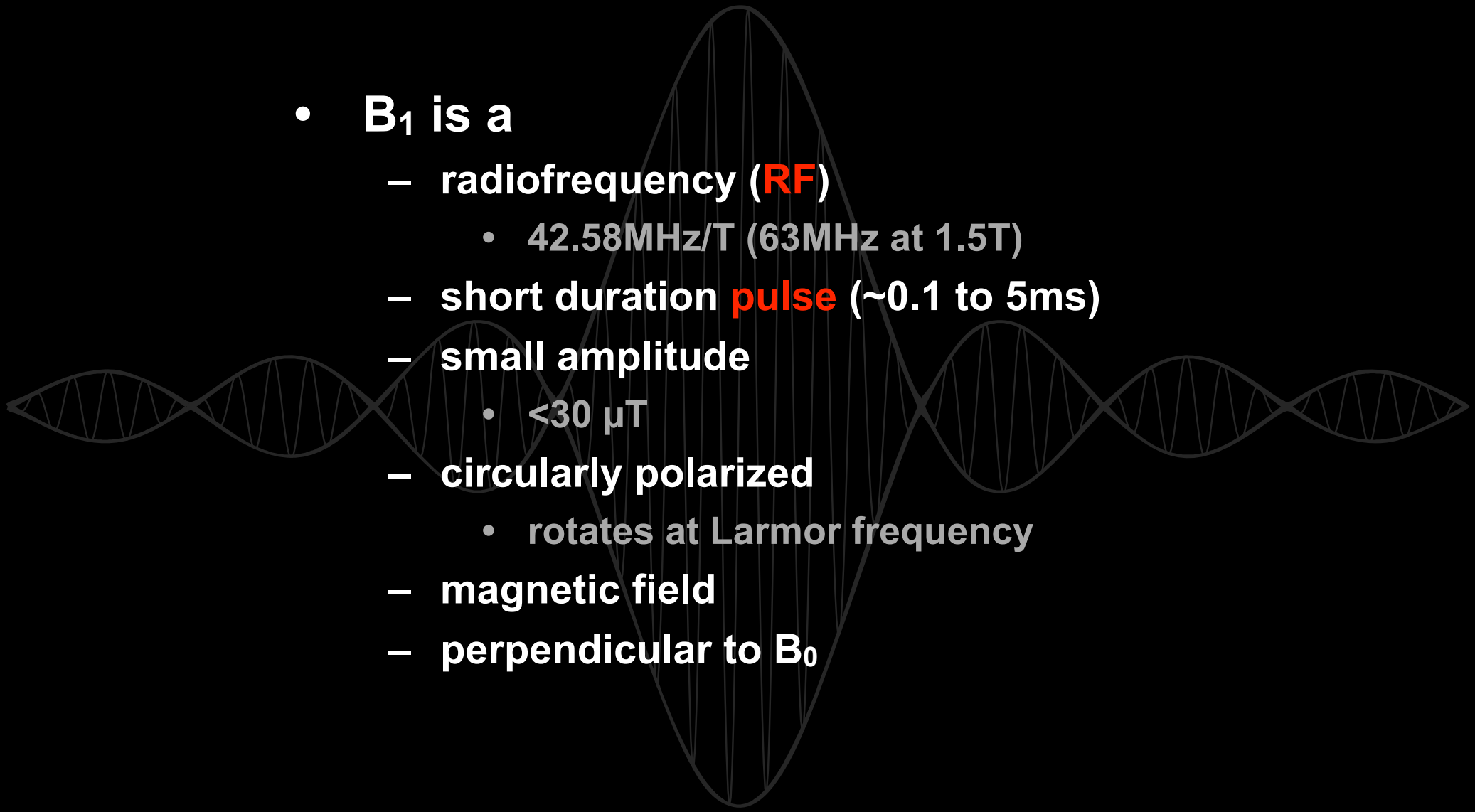
B_1

Coil

Gradients

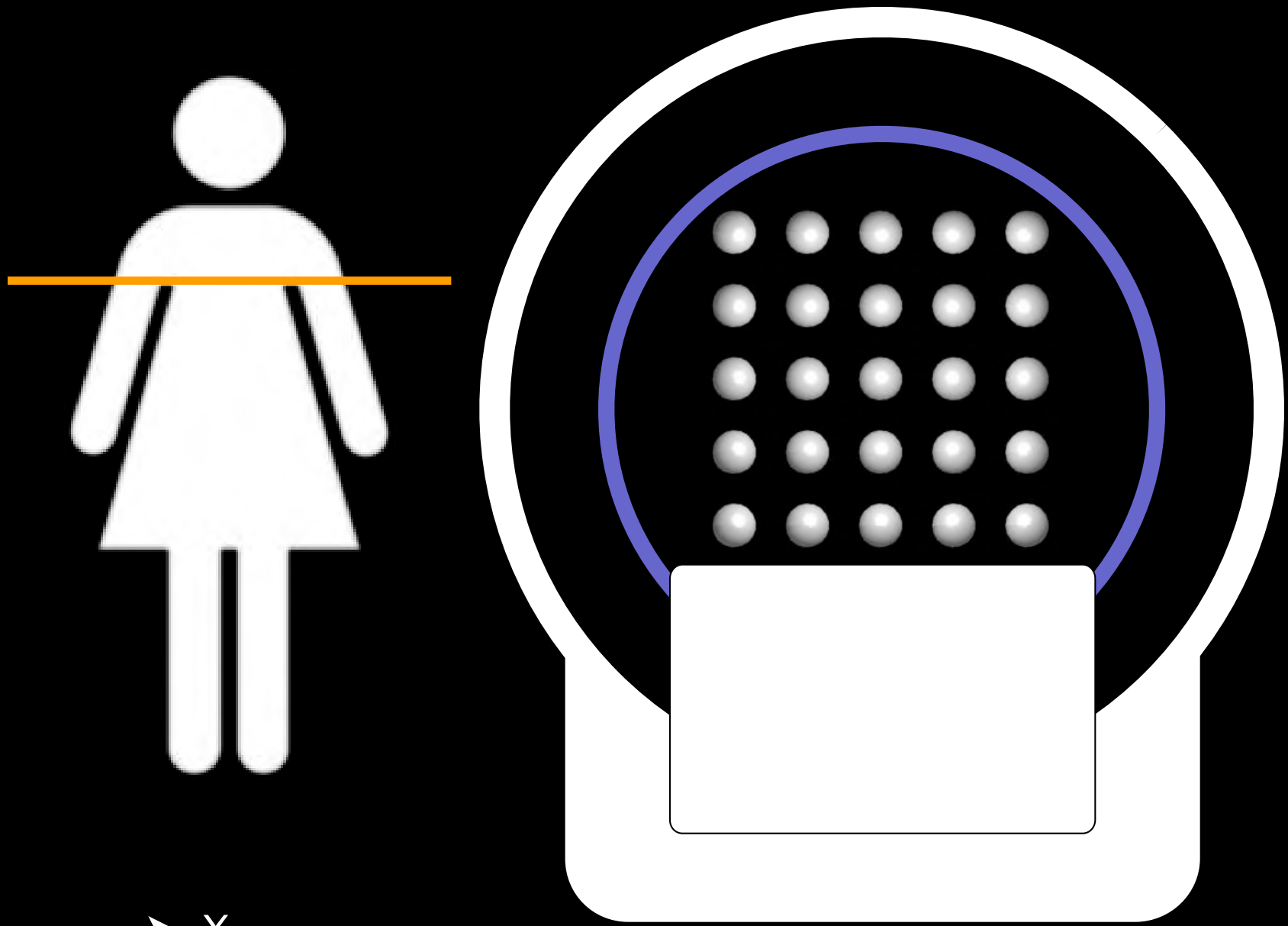
FFT

B_1 Field - RF Pulse



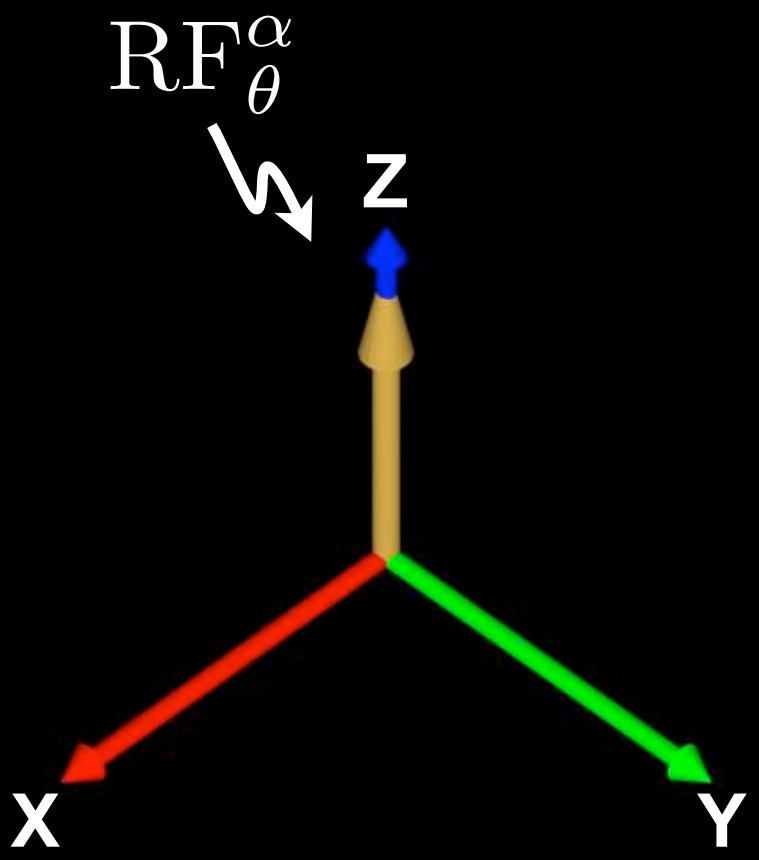
- B_1 is a
  - radiofrequency (RF)
    - 42.58MHz/T (63MHz at 1.5T)
  - short duration pulse (~0.1 to 5ms)
  - small amplitude
    - <30 μT
  - circularly polarized
    - rotates at Larmor frequency
  - magnetic field
  - perpendicular to B_0

RF Excitation

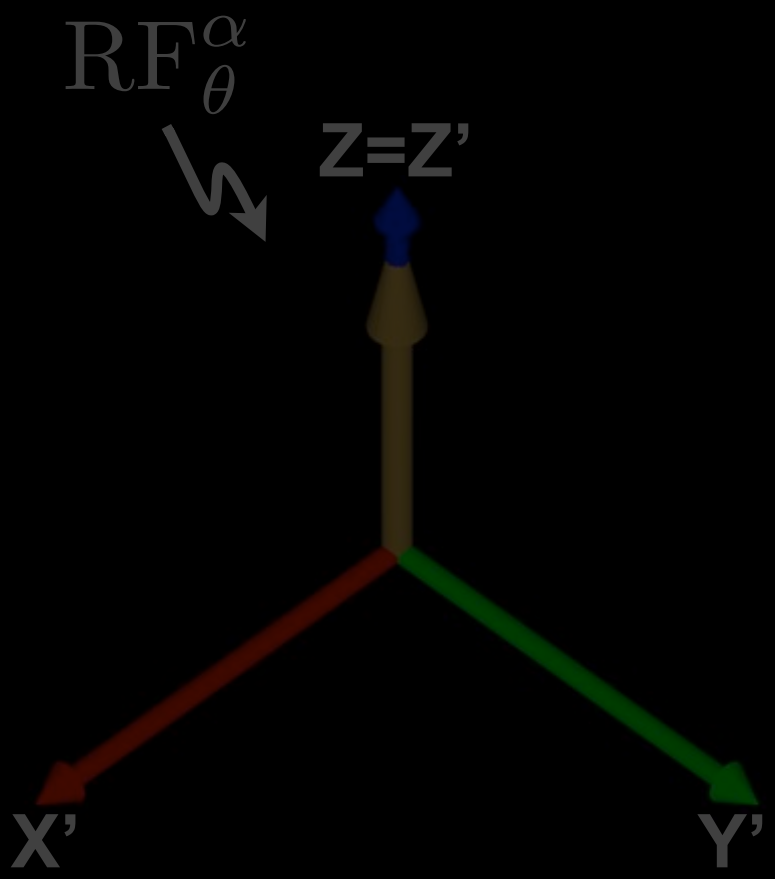


RF pulses generate transverse magnetization for a *slice*.

Lab vs. Rotating Frame



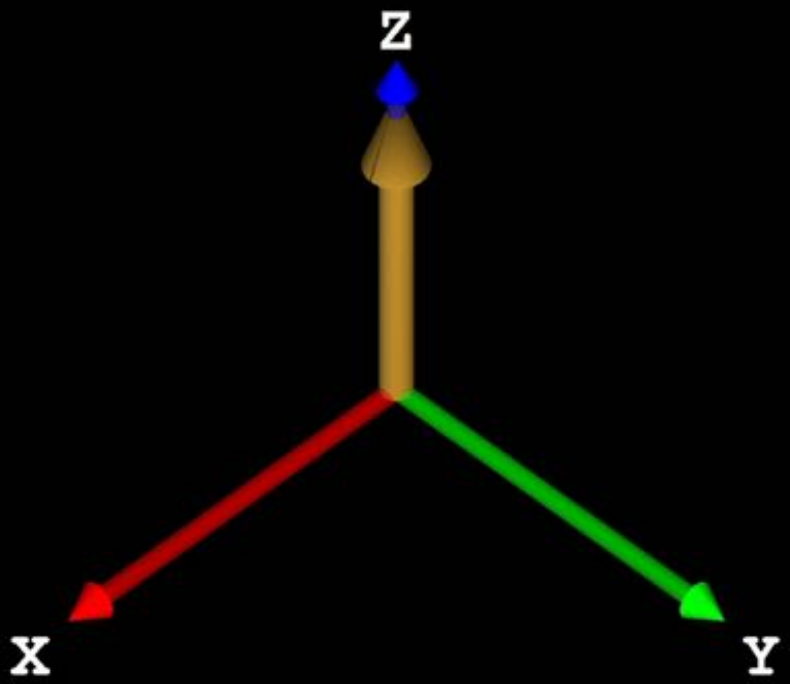
Laboratory Frame



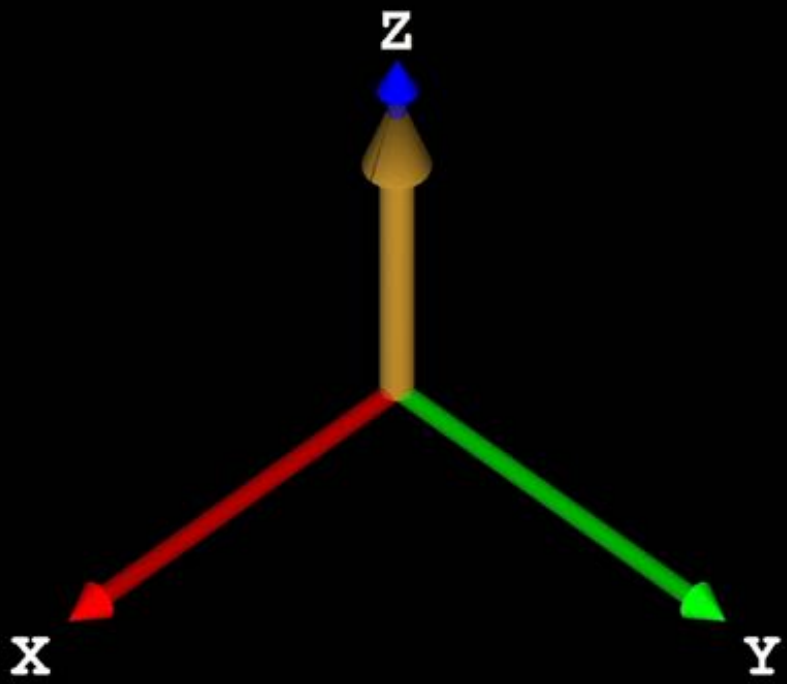
Rotating Frame

A lot of the math can be done more easily in the rotating frame.

RF Pulses



Small Flip Angle



Inversion Pulse

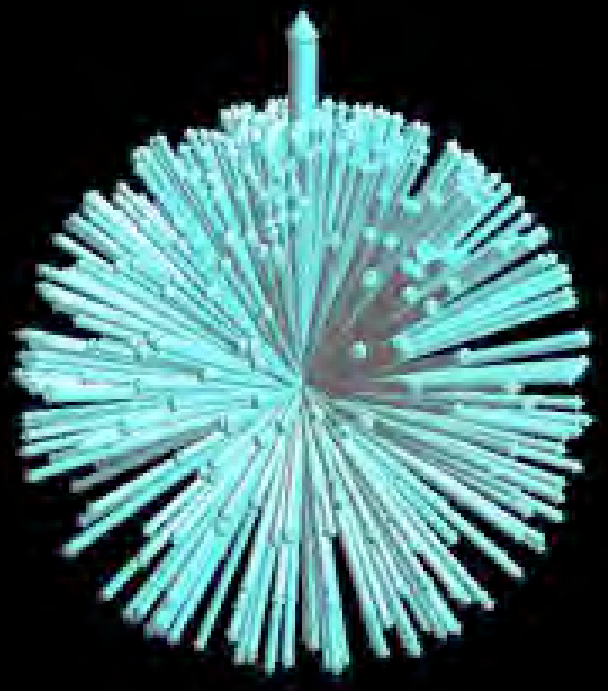
Resonance

"Establishment of a phase coherence among these 'randomly' precessing spins in a magnetized spin system is referred to as *resonance*."

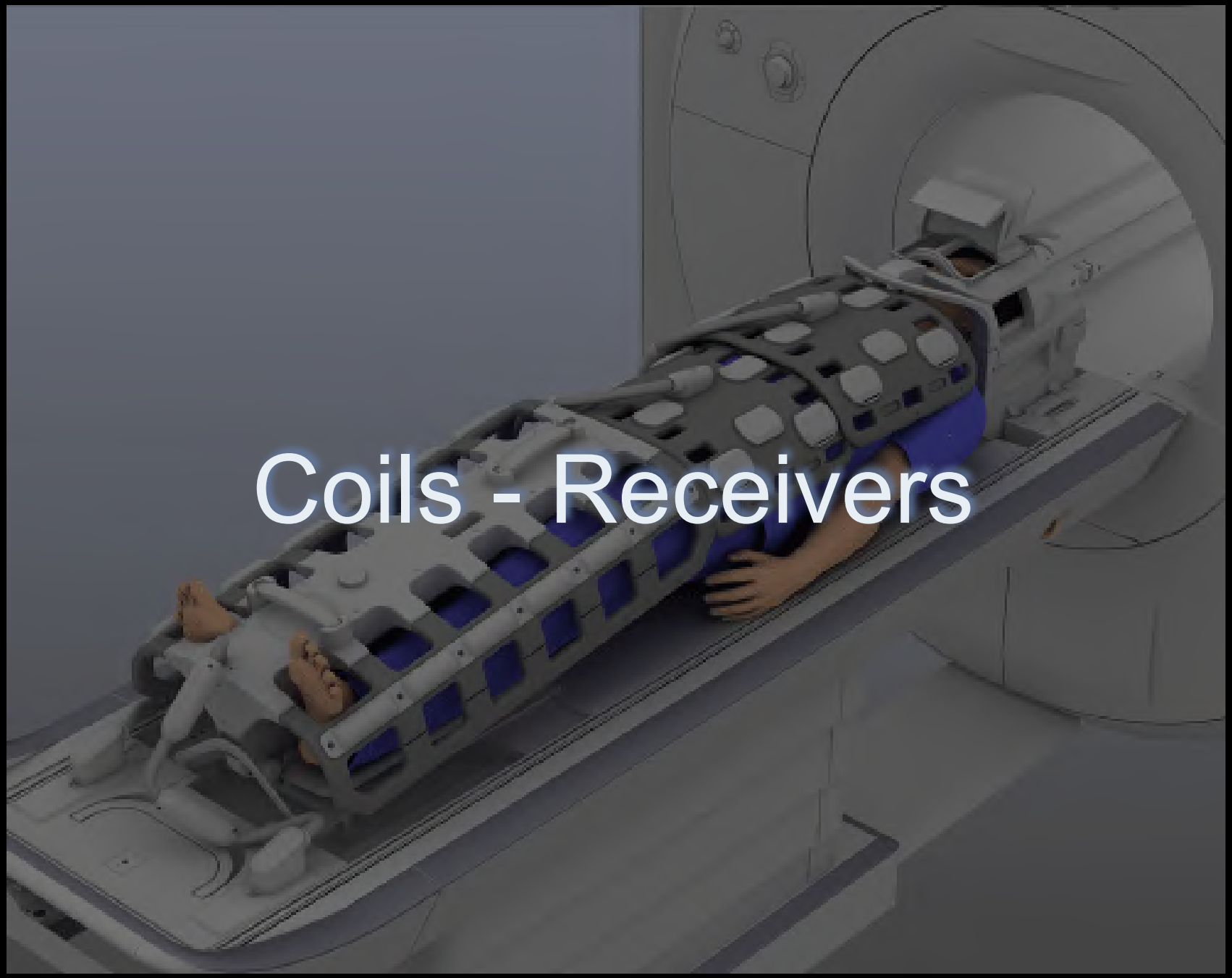
– Liang & Lauterbur p.69

$\vec{B}$

$N_{\uparrow} \approx N_{total} \times (1 + 2.25 \times 10^{-6})$



$N_{\downarrow} \approx N_{total} \times (1 - 2.25 \times 10^{-6})$


Coils - Receivers

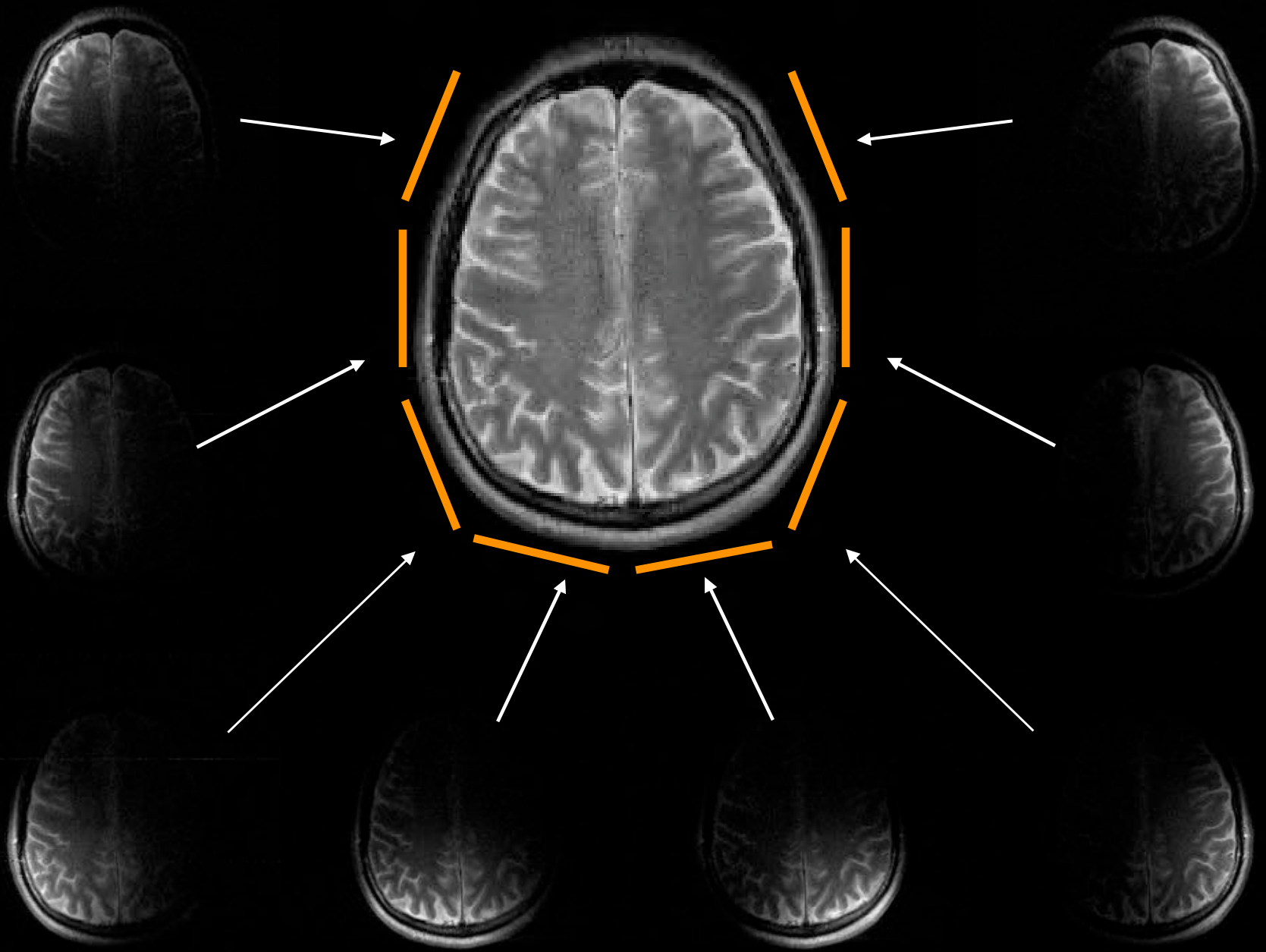
Dipoles to Images

$\vec{\mu}$	Magnetic Moment	B_0
$\downarrow$		
$\vec{M}_z$	Bulk Magnetization	B_1
$\downarrow$		
$\vec{M}_{xy}$	Transverse Magnetization	Coil
$\downarrow$		
$S(t)$	Received Signal	
$\downarrow$		Gradients
$S\left(\vec{k}\right)$	k-space signal	
$\downarrow$		FFT
$I(\vec{x})$	Image	

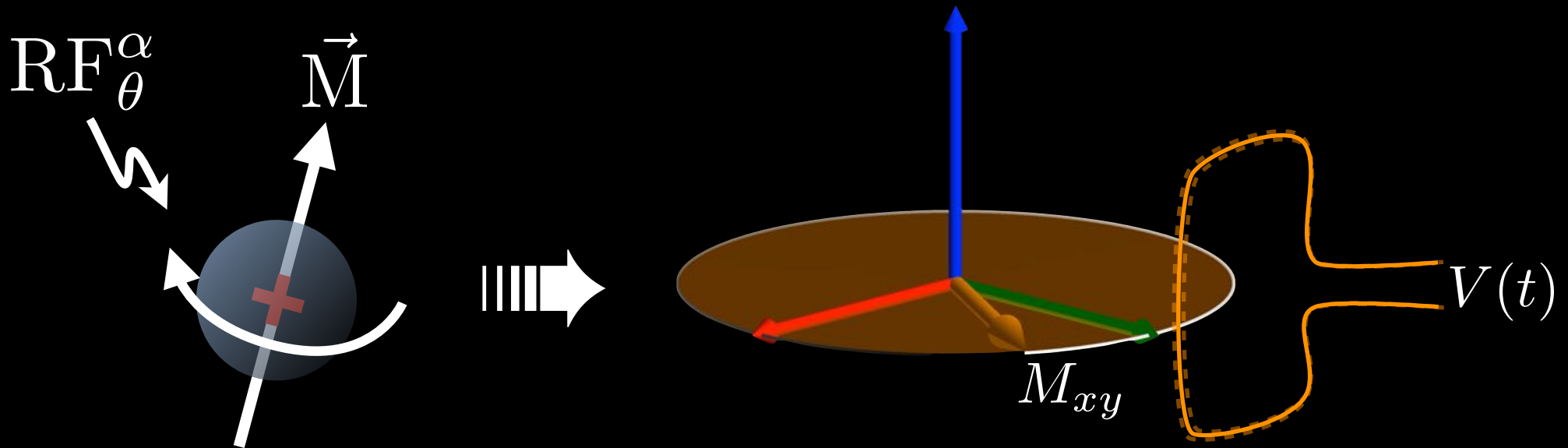
Coils

8-Channel Head Coil

Each coil element has a unique sensitivity profile.

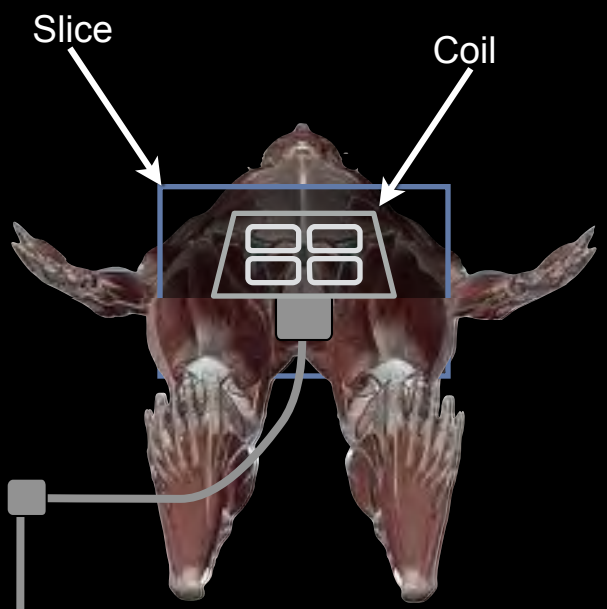


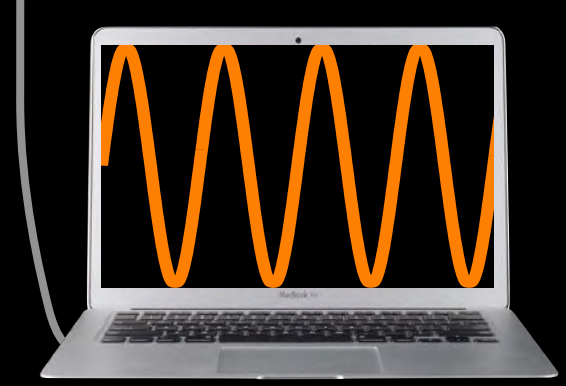
Signal Reception

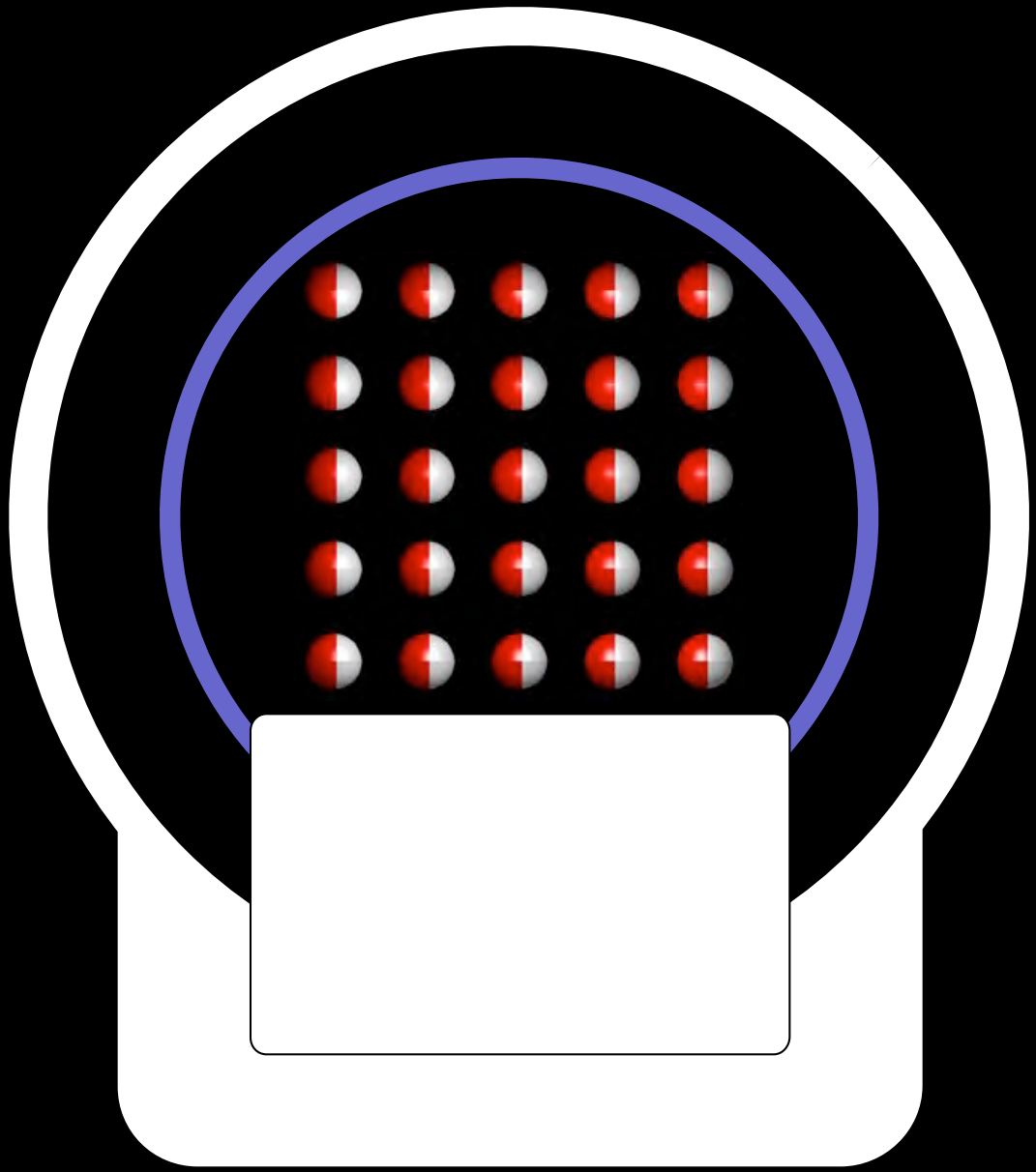


Faraday's Law of Induction

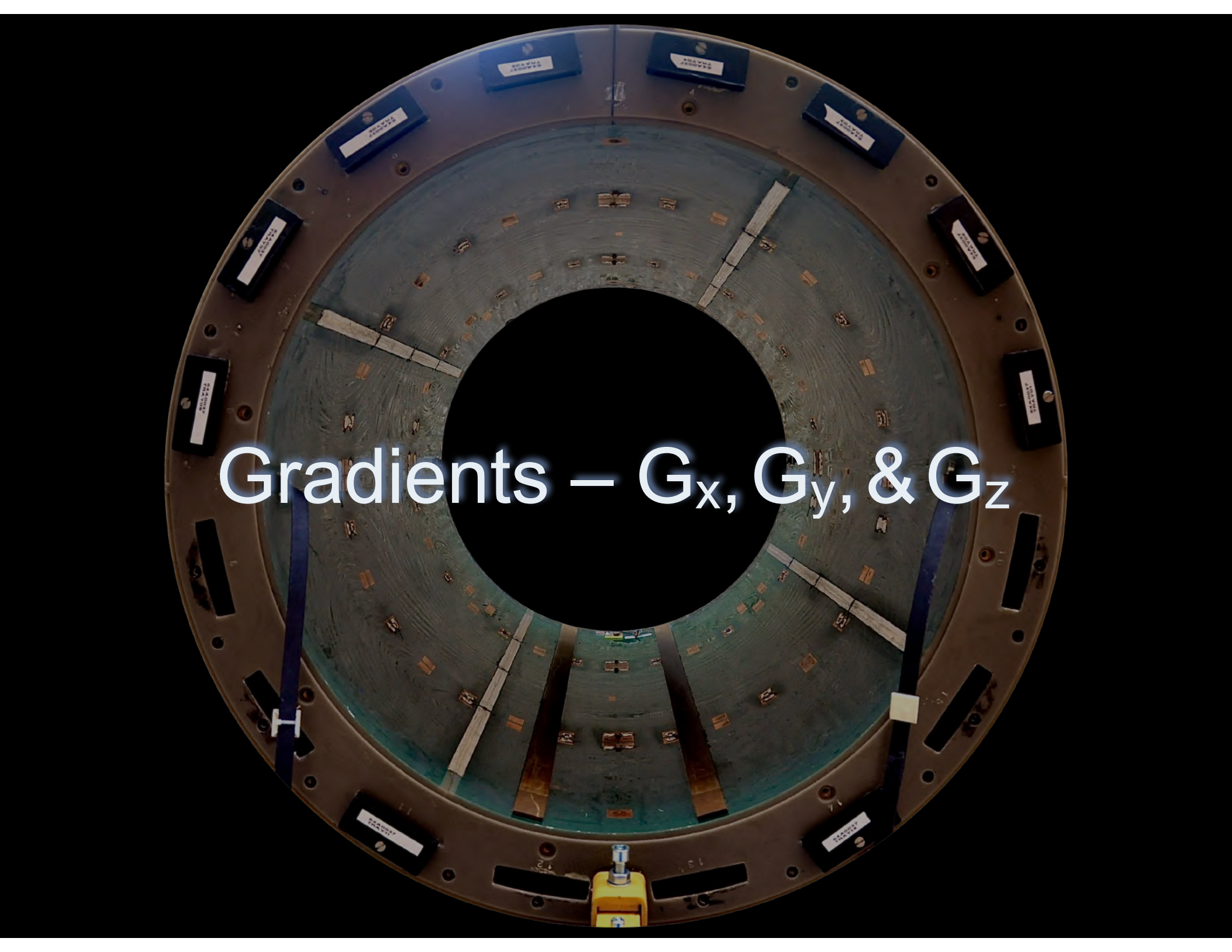
Faraday's Law of Induction







The trick is to encode spatial information and image contrast in the echo.


Gradients – G_x, G_y, & G_z

Dipoles to Images

$\vec{\mu}$	Magnetic Moment	B_0
$\downarrow$		
$\vec{M}_z$	Bulk Magnetization	B_1
$\downarrow$		
$\vec{M}_{xy}$	Transverse Magnetization	Coil
$\downarrow$		
$S(t)$	Received Signal	Gradients
$\downarrow$		
$S\left(\vec{k}\right)$	k-space signal	FFT
$\downarrow$		
$I\left(\vec{x}\right)$	Image	

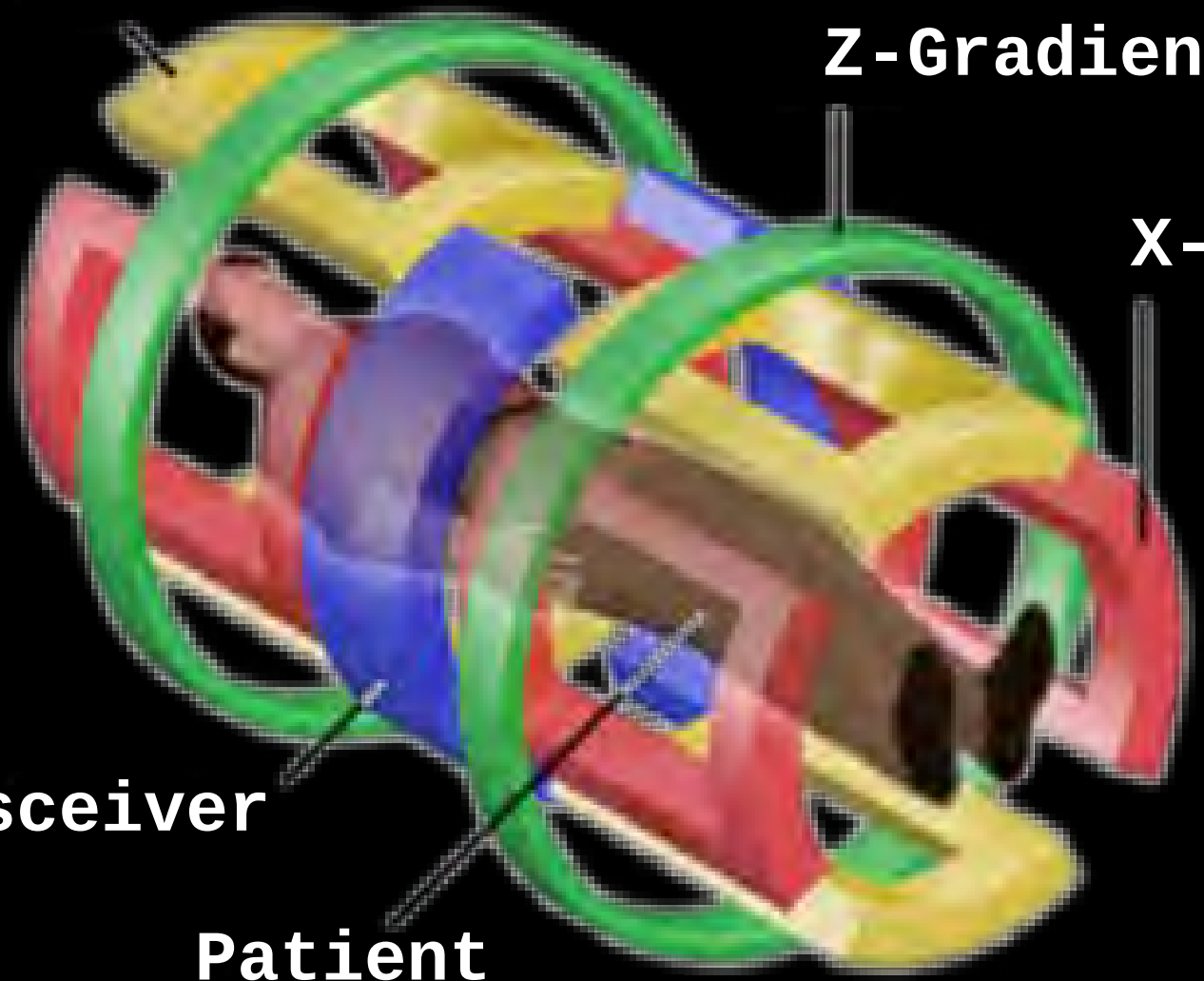
Gradients

- Primary function
  - Encode spatial information
    - Slice selection
    - Phase encoding
    - Frequency encoding
- Secondary functions
  - Sensitize/de-sensitize images to motion
  - Minimize artifacts (crushers & spoilers)
  - Magnetization re-phasing in slice selection
  - Magnetization de-phasing during readout

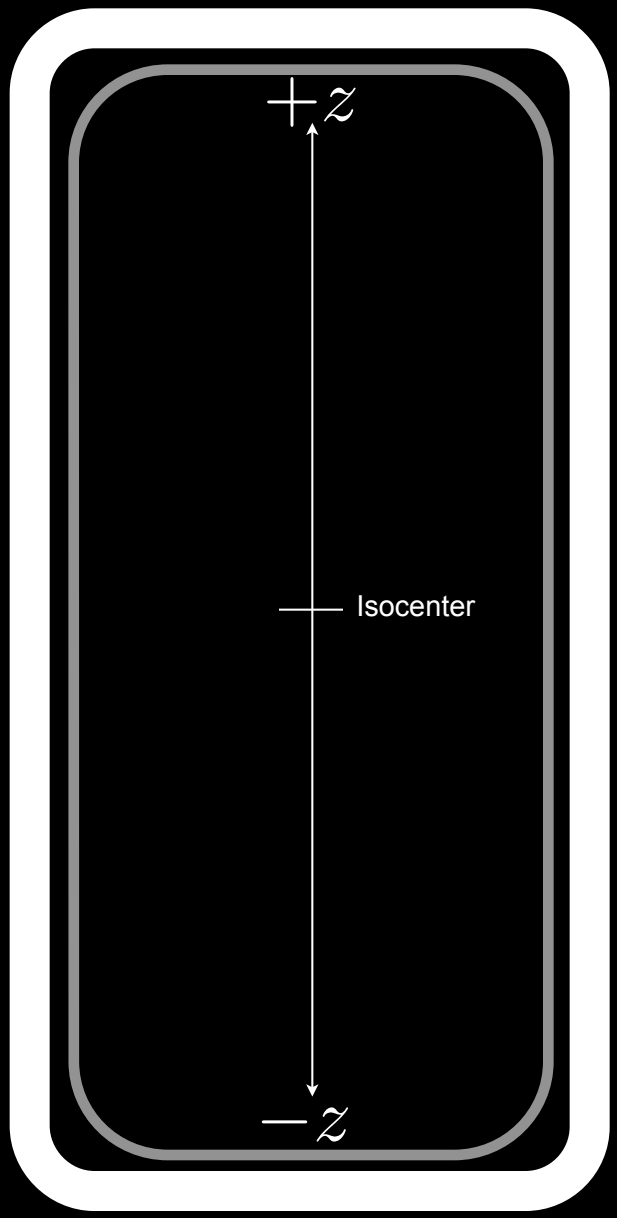
Gradients

- Gradients are a:
  - **Small**
    - <5G/cm (<0.0075T @ edge of 30cm FOV)
  - **Spatially varying**
    - Linear gradients
    - Adds to B_0 only in Z-direction
  - **Time varying**
    - Slewrate Max. ~150-200mT/m/ms
  - **Magnetic fields**
    - Adds/Subtracts to the B_0 field
  - **Parallel to B_0**

MRI Instrumentation



No Gradients Turned On



B_0

B_0

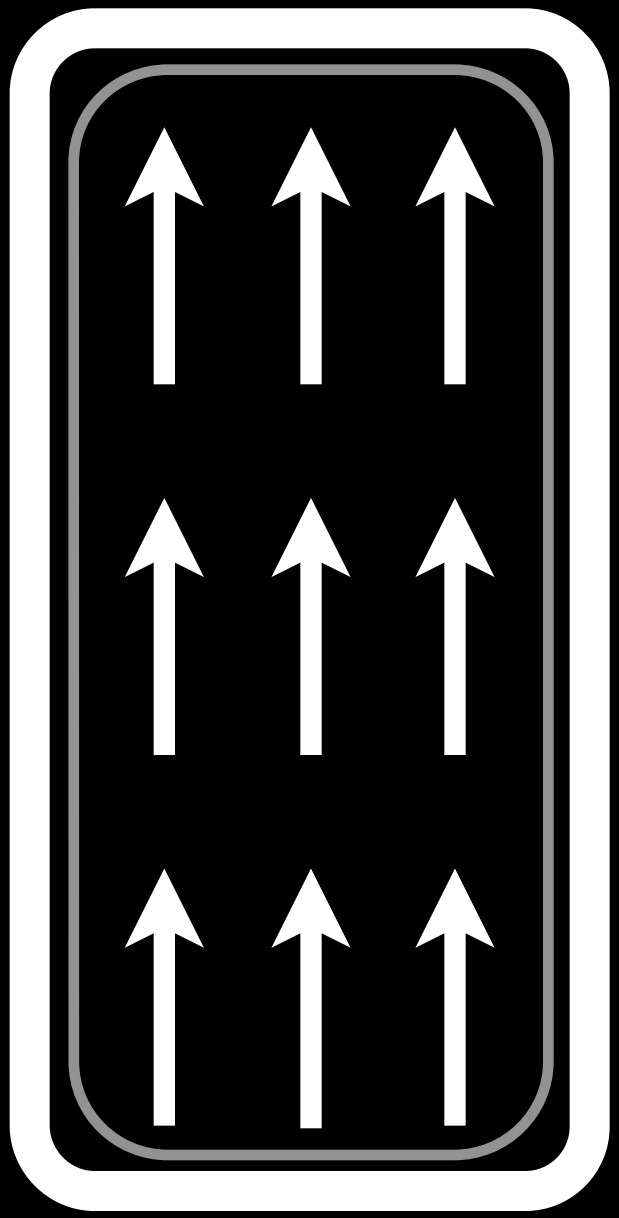
B_0

z

x

$$\omega = \gamma B_0$$

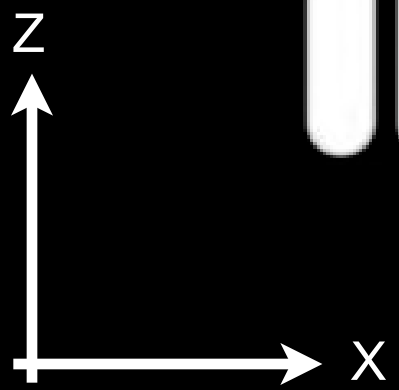
No Gradients Turned On



B_0

B_0

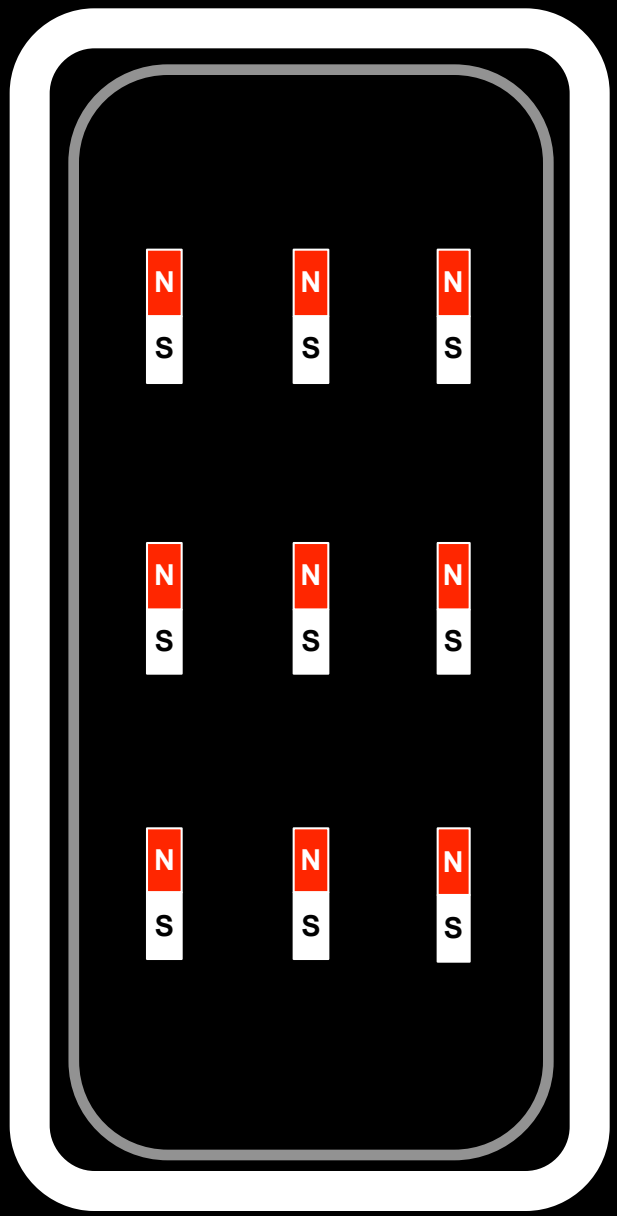
B_0



$$\omega = \gamma B_0$$

Length of arrow indicates strength of local field.

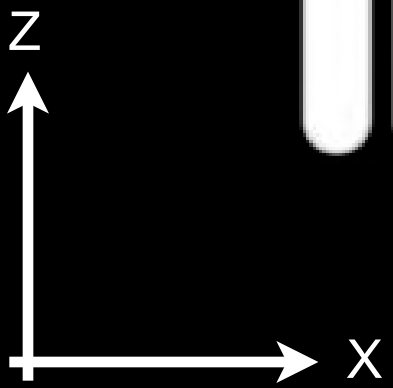
No Gradients Turned On



B_0

B_0

B_0



$$\omega = \gamma B_0$$

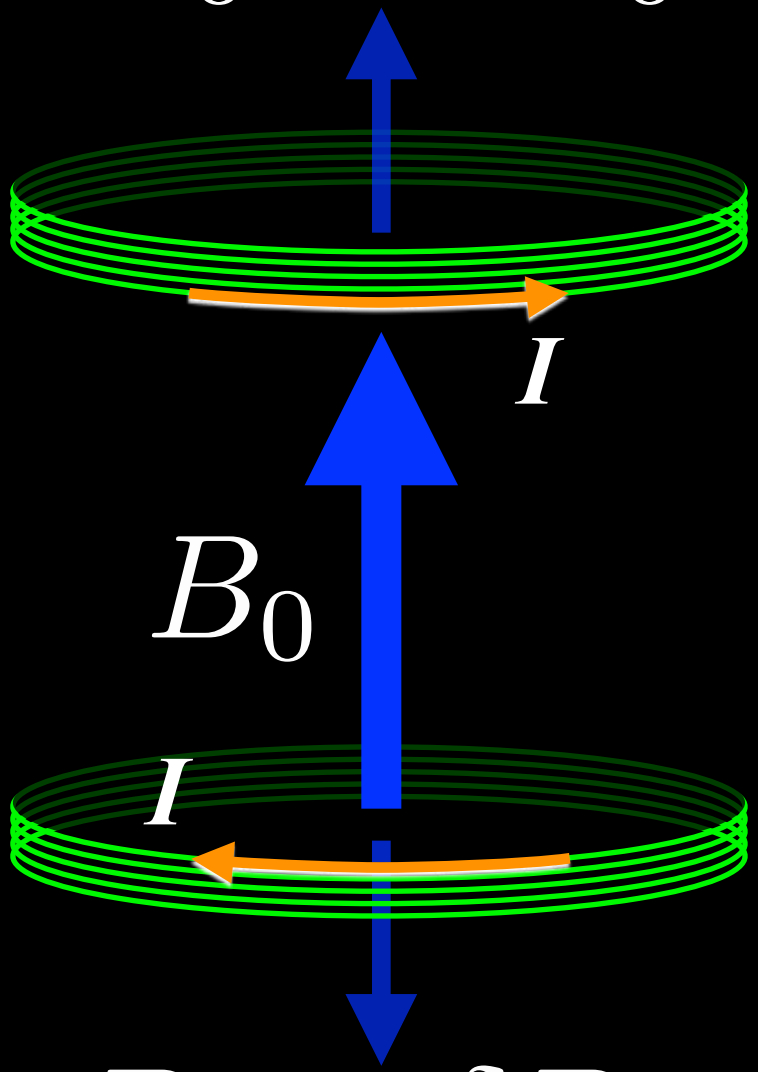
Everything precesses at the Larmor frequency.

Z Gradients

$$B_0 + \delta B_0$$

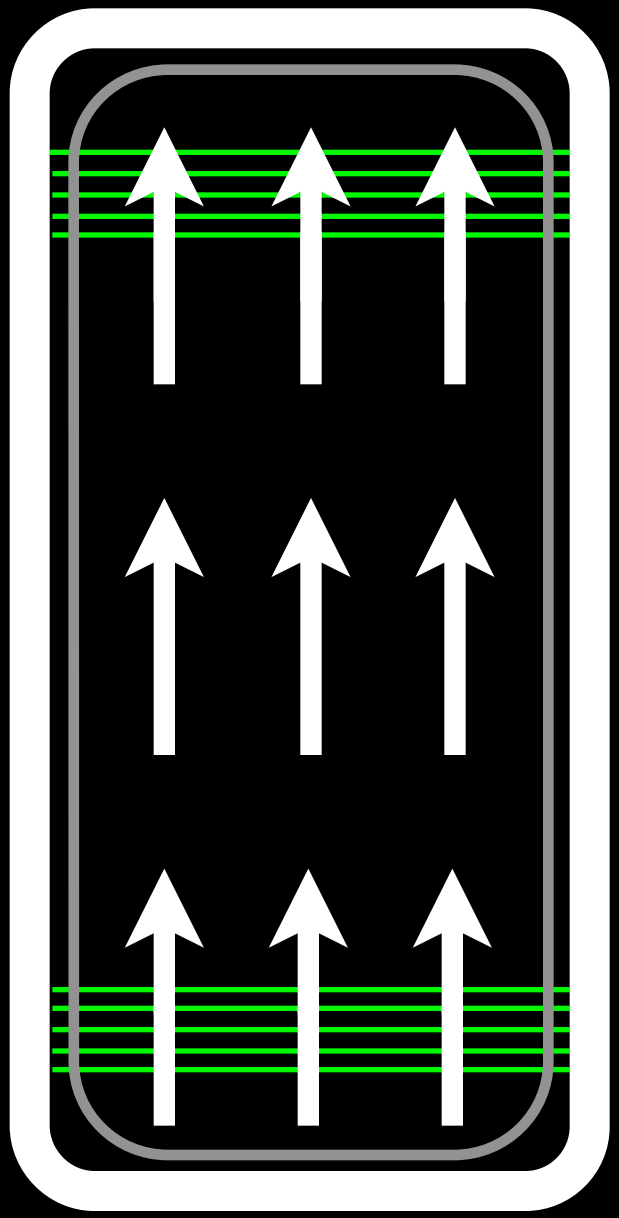
$$\delta B_0 = B_{G,z}(z) = G_z \cdot z$$

Maxwell Pair Coil



$$B_0 - \delta B_0$$

Z-Gradients



$B_0 + \delta B_0$

B_0

$B_0 - \delta B_0$

z

x

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Length of arrow indicates strength of local field.

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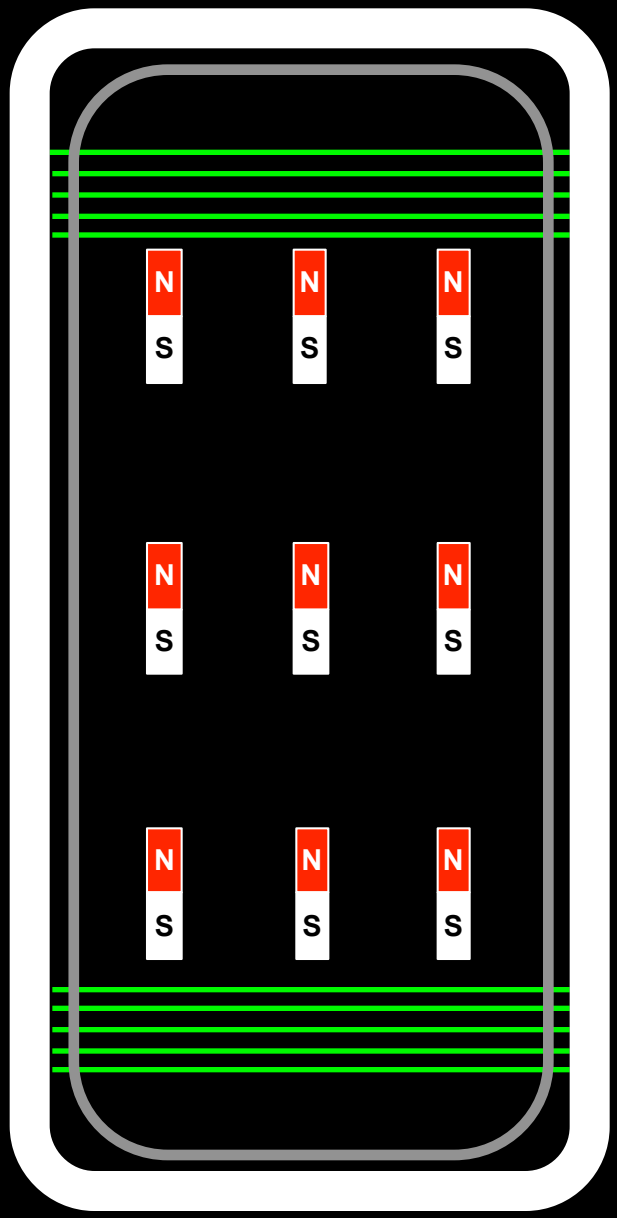
Z-Gradients



$$B_0 + \delta B_0$$

$$B_0$$

$$B_0 - \delta B_0$$

z

x

$$\omega = \gamma\left(B_0 + G_z \cdot z\right)$$

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Gradients give rise to isochromats (planes of common frequency).

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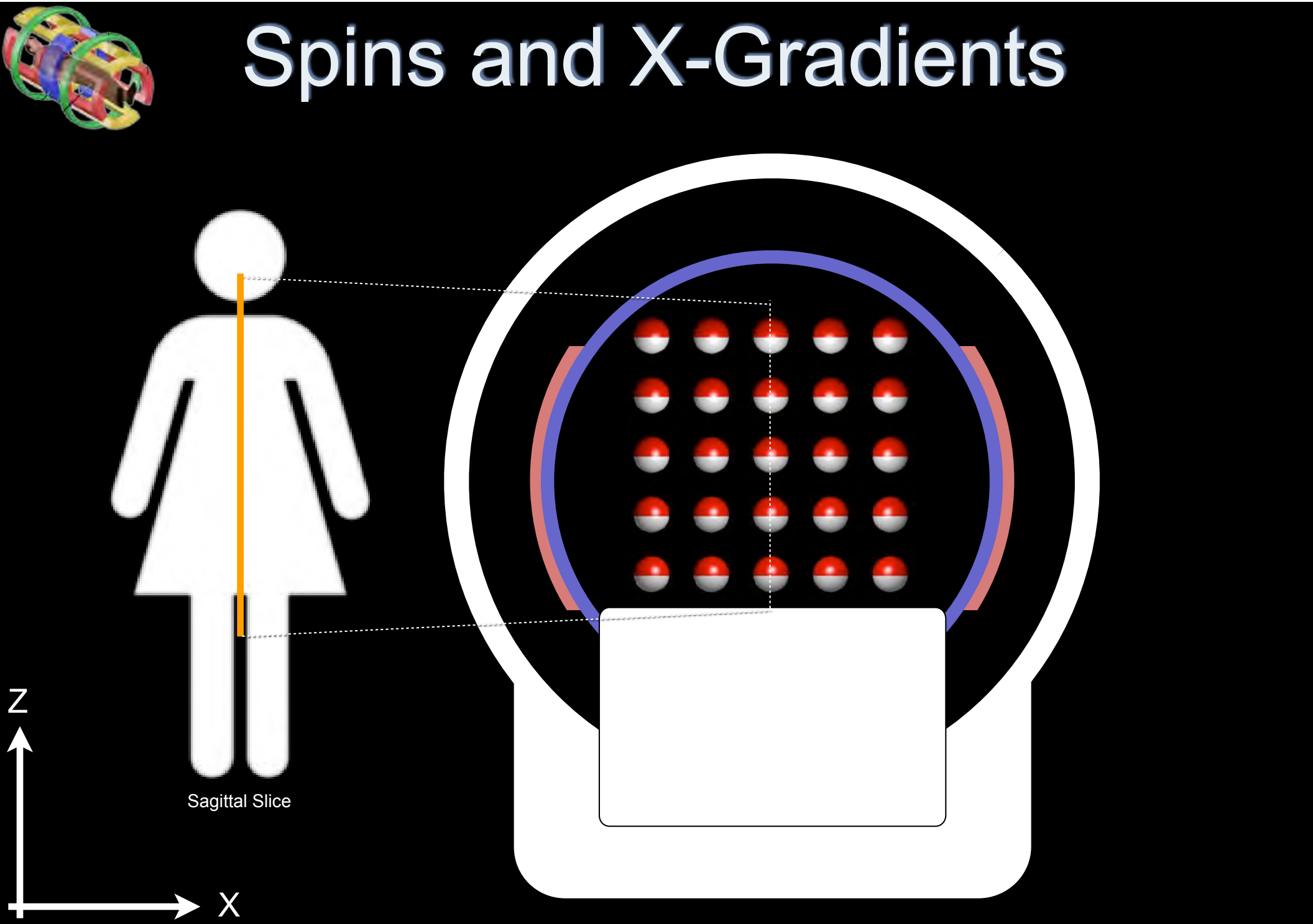
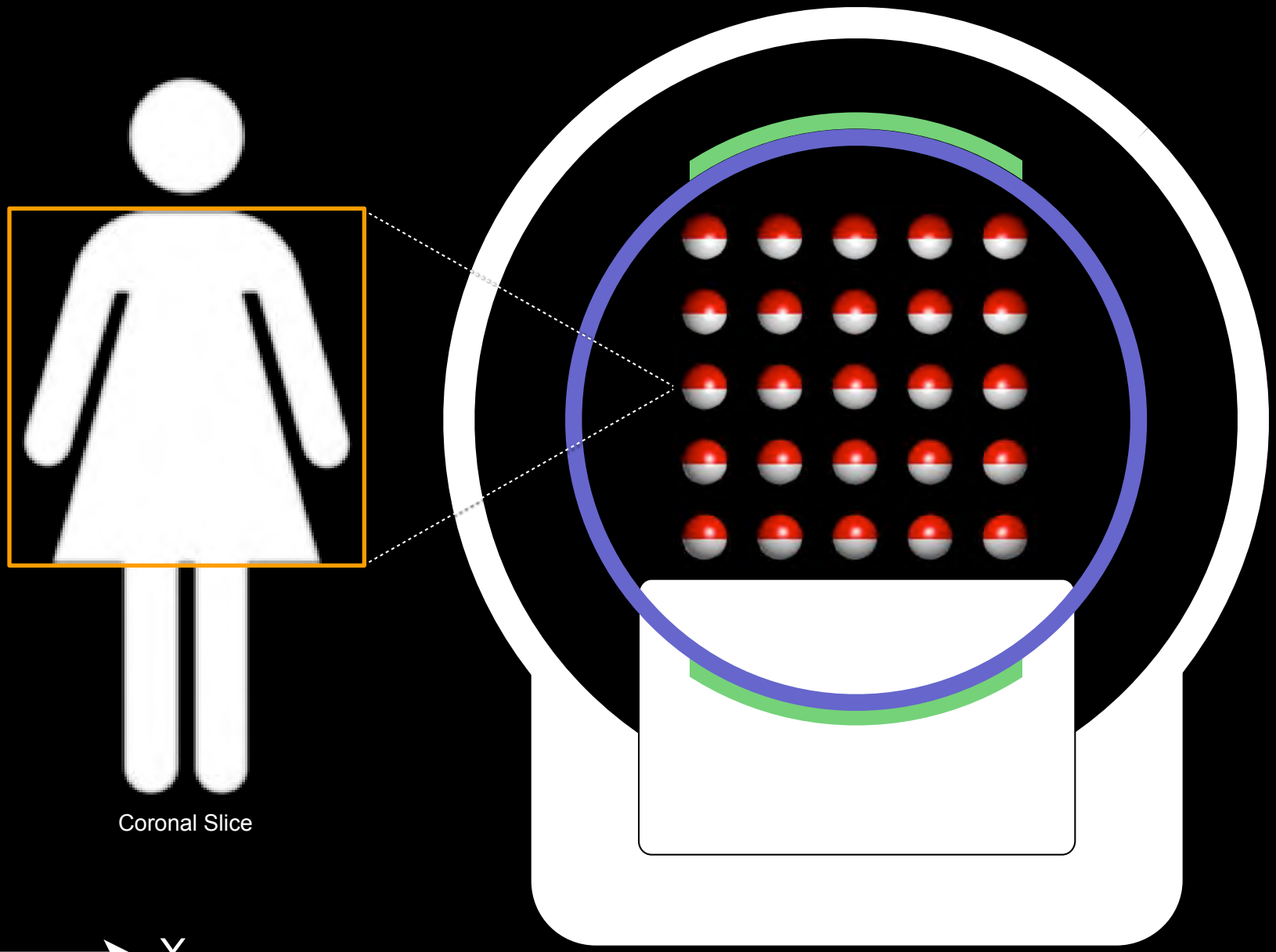
Spins and X-Gradients



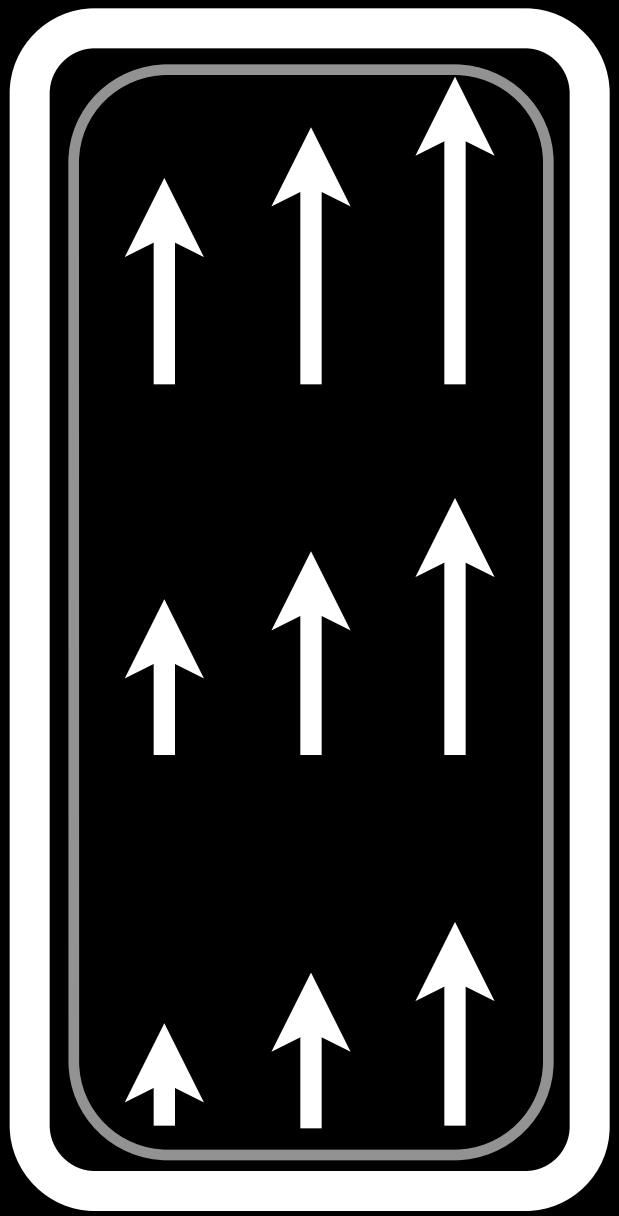
Gradients give rise to isochromats (planes of common frequency).

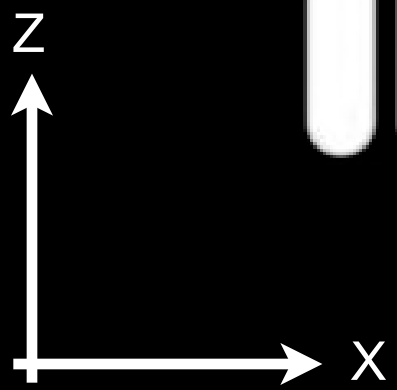
Spins and Y-Gradients



Gradients add/subtract to B_0 along a specific direction.

How do we do this?

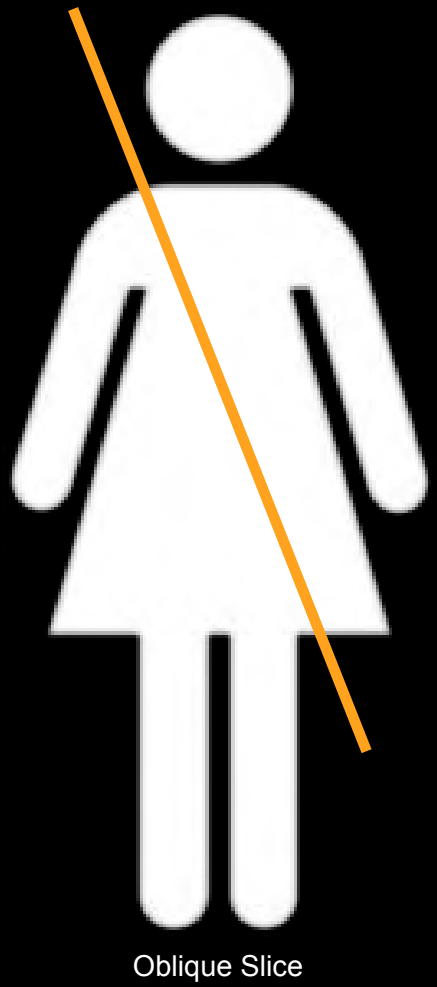




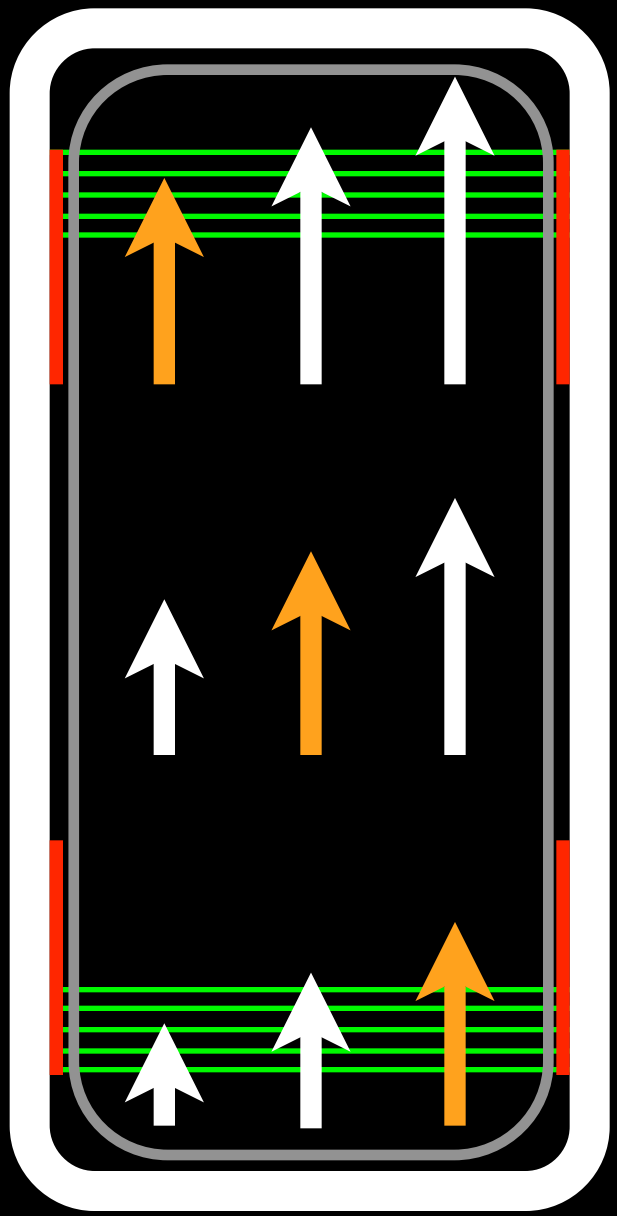
Gradients can be combined.

X+Z-Gradients

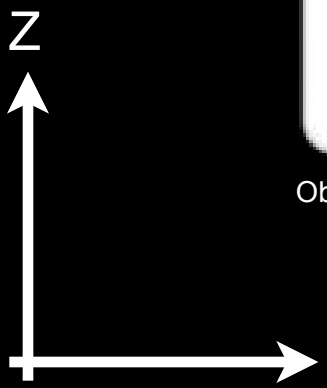
Possible Slice



Oblique Slice



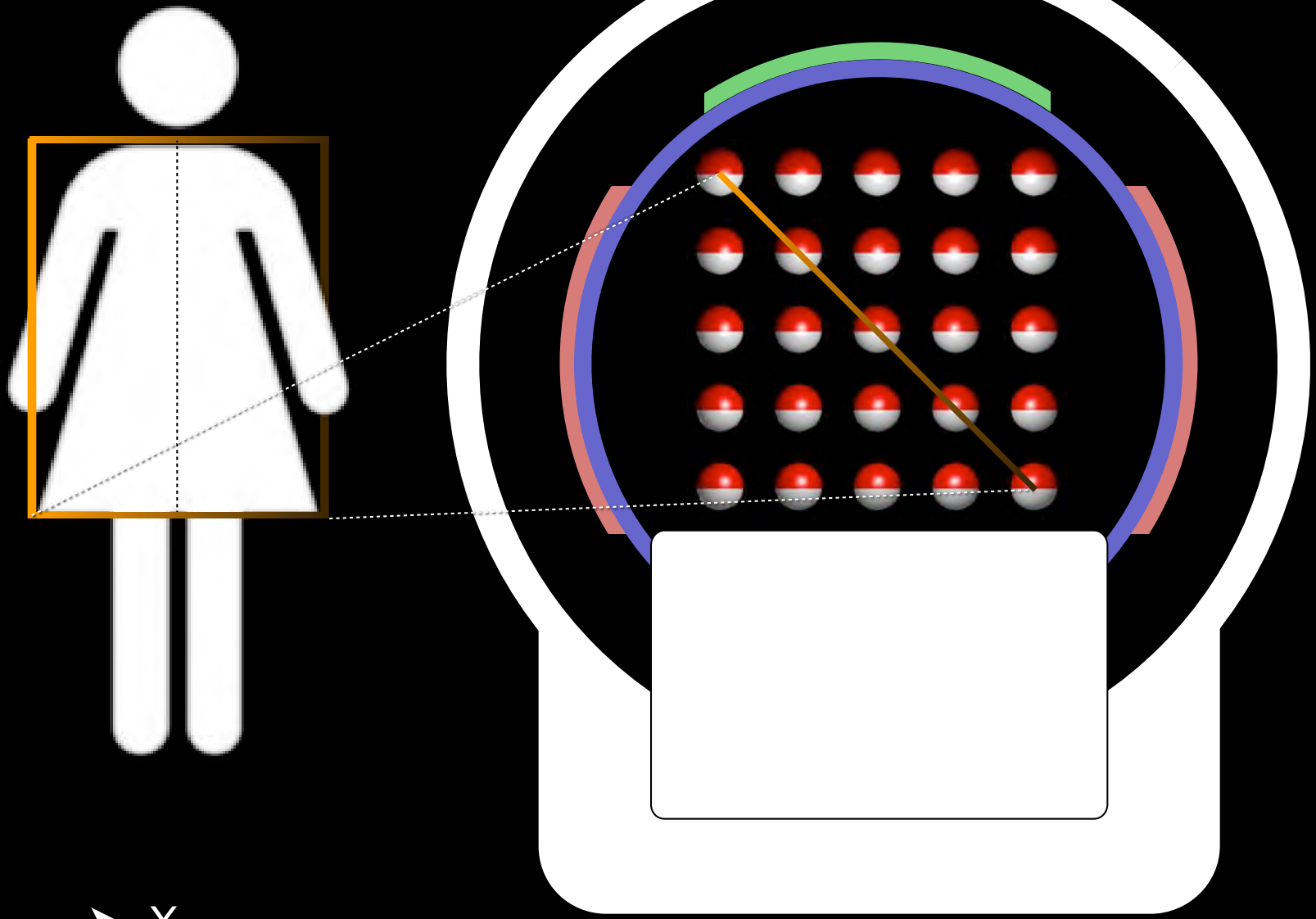
Spin Isochromat



$$\omega = \gamma B_0$$

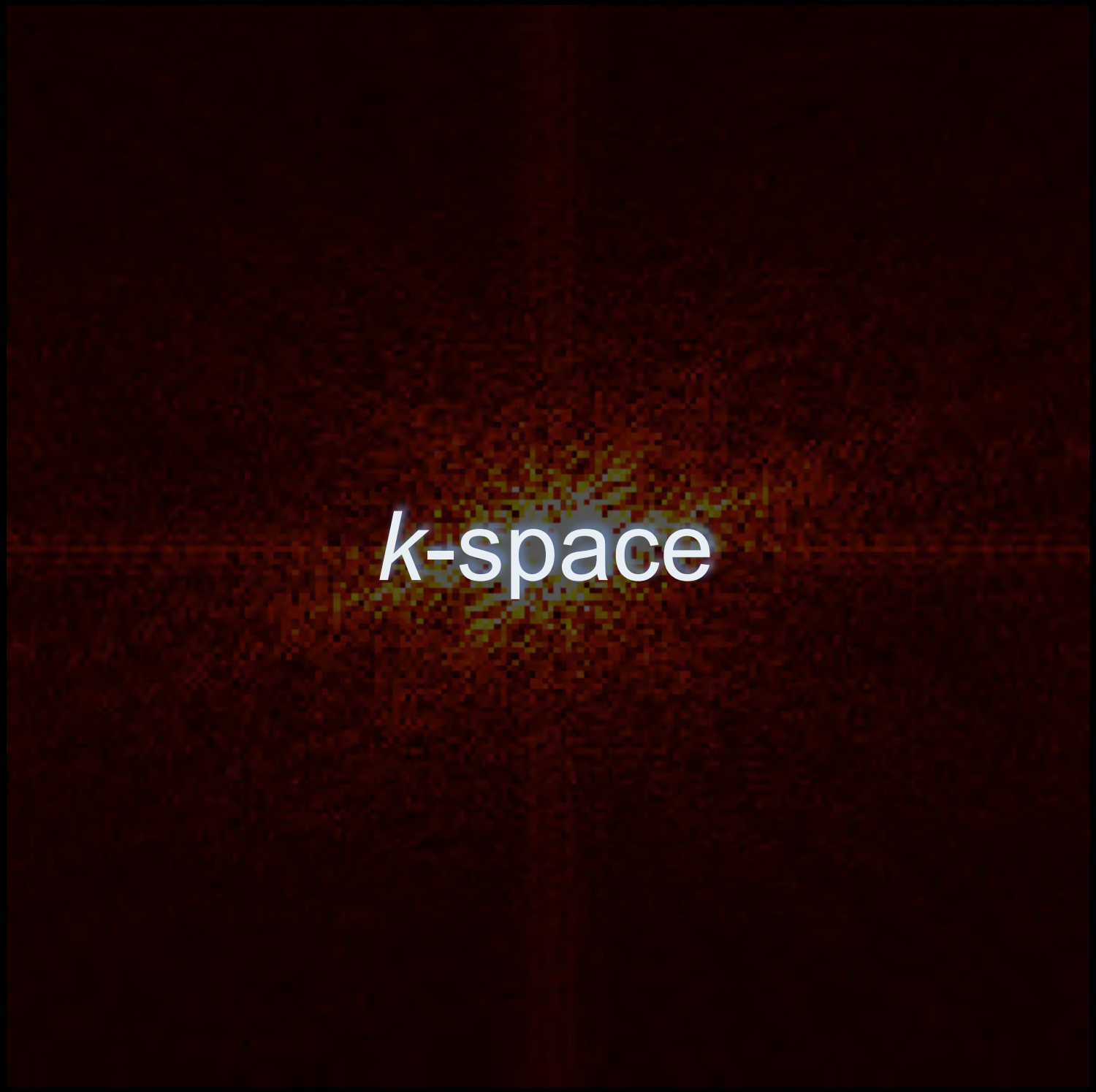
Gradients combined with RF pulses produce slice selection.

Spins and X- & Y-Gradients



Simultaneously gradients create an arbitrary isochromat plane.

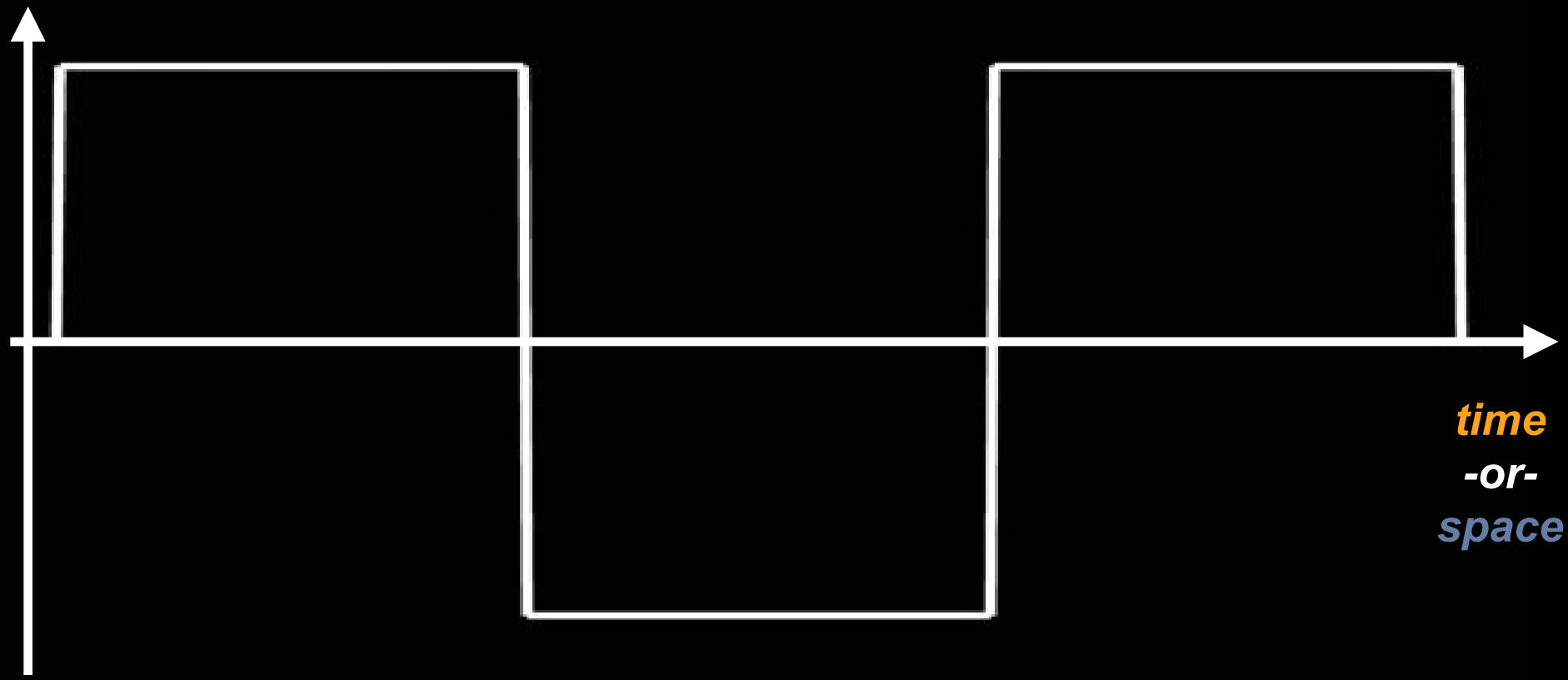
Gradients DO NOT do this!


k-space

What is *k*-space?

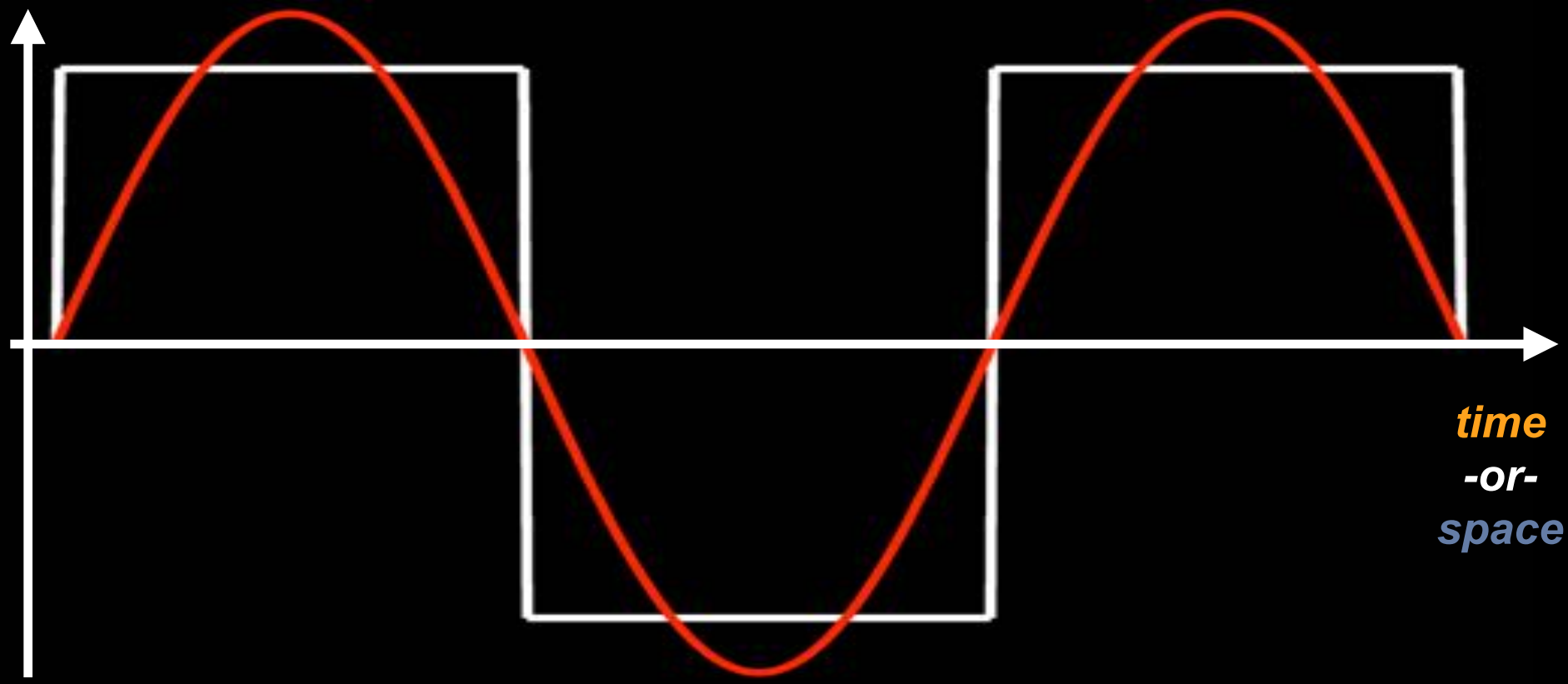
- ***k*-space is the raw data collected by the scanner.**
  - A point in k-space tells us about the presence/absence of a spatial frequency (pattern) in the acquired image.
  - Each echo measures *many* of the spatial frequencies that comprise the object.
  - *k*-space has units of cm^{-1} or mm^{-1}
    - Audio signals have units of Hertz (s^{-1})
- **Gradients**
  - Help extract spatial frequency information
  - Move us around in k-space
- **A line of *k*-space is filled by an echo**
- **2D FT of *k*-space produces the image**

1D *k*-space



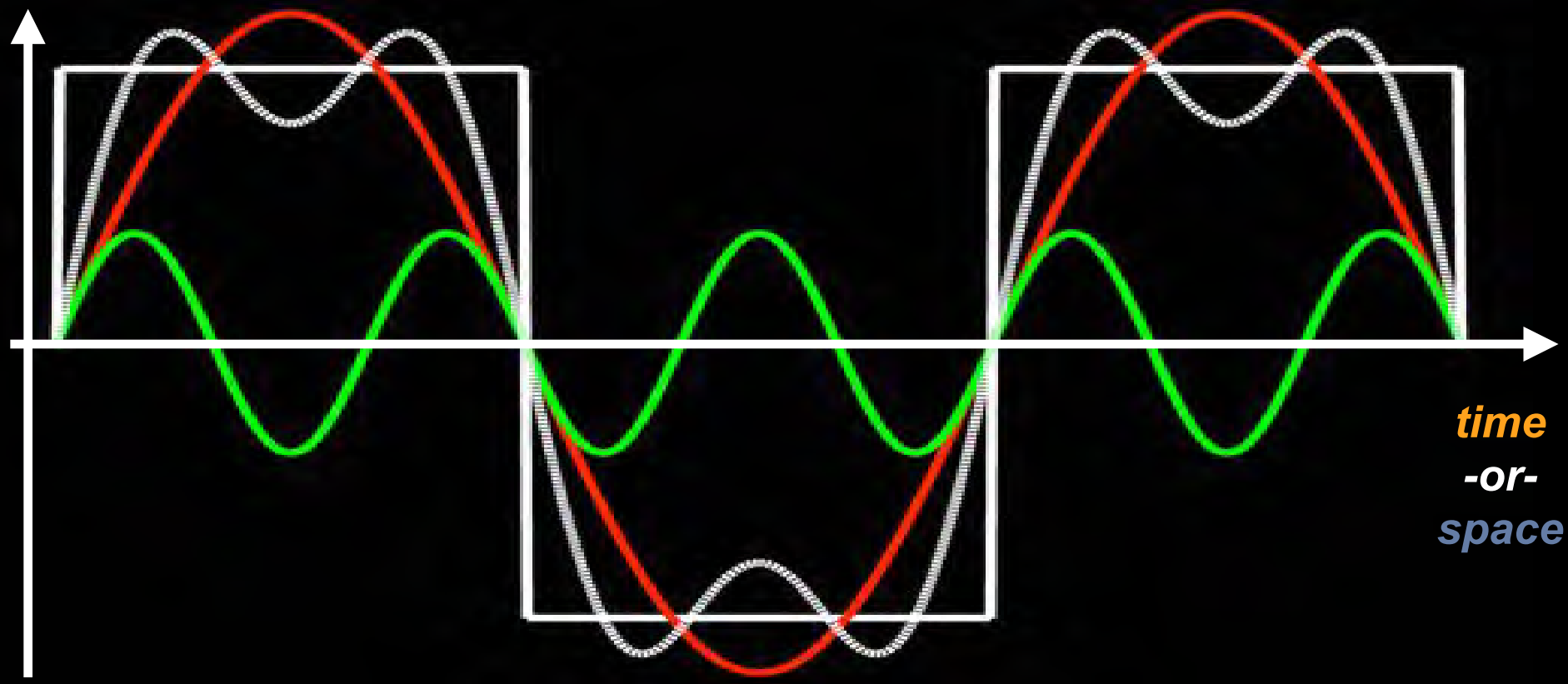
Any signal/image can be decomposed into a summation of sine waves of appropriate amplitude.

1D *k*-space



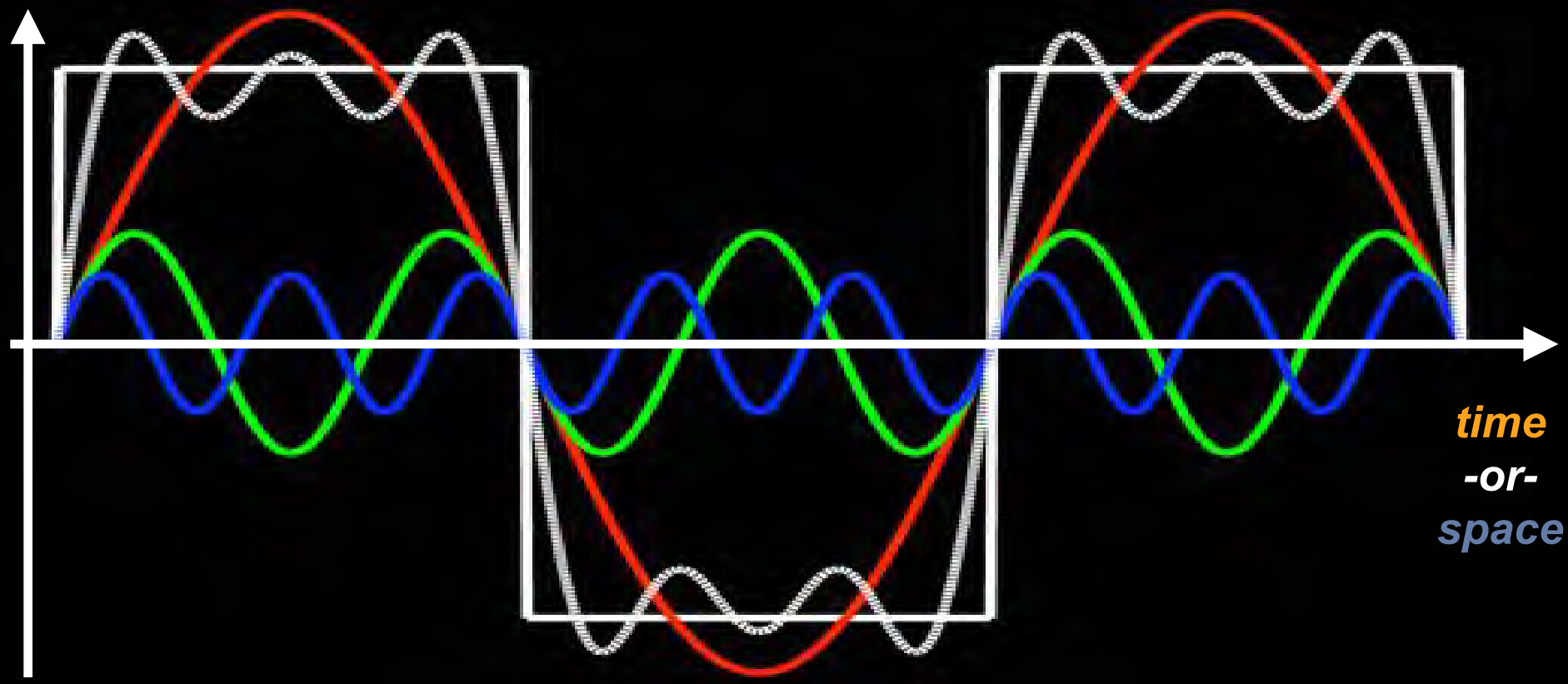
Any signal/image can be decomposed into a summation of sine waves of appropriate amplitude.

1D *k*-space



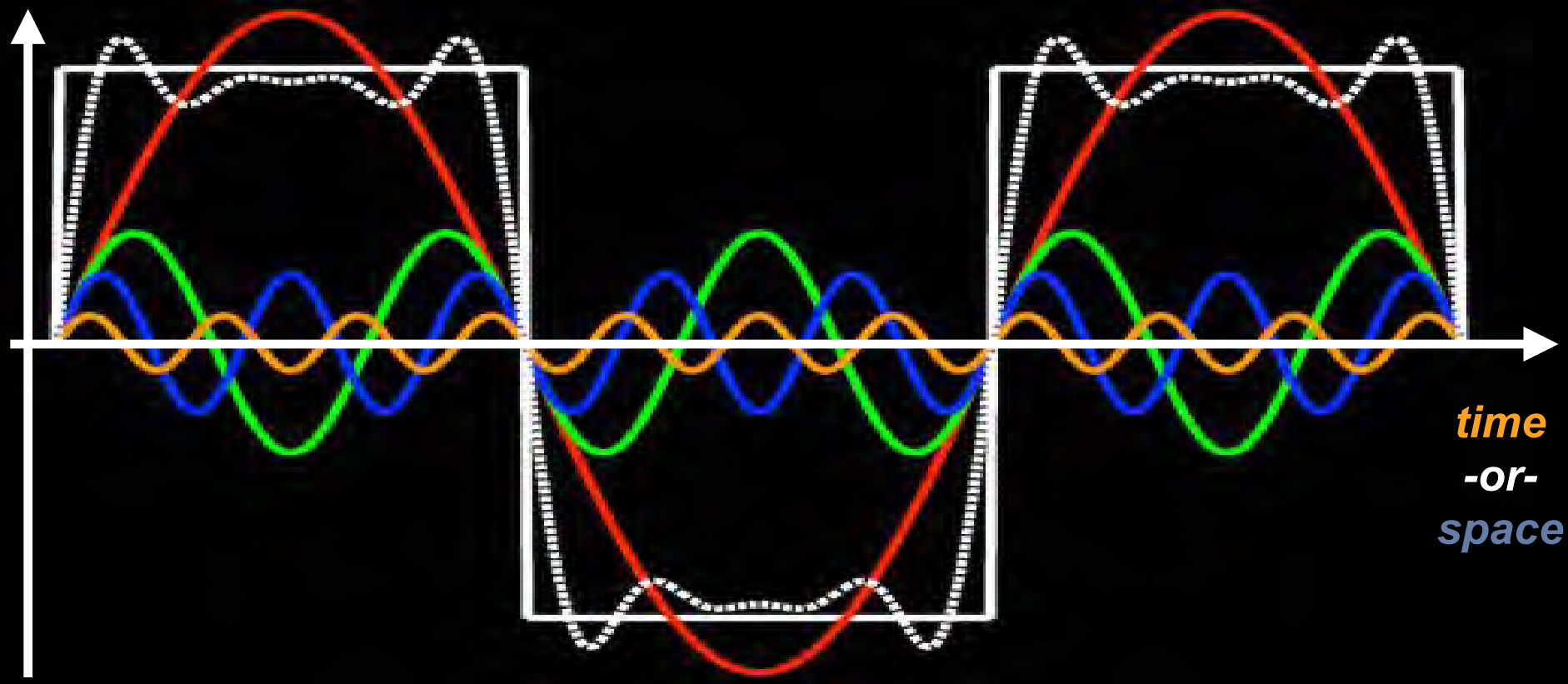
Any signal/image can be decomposed into a summation of sine waves of appropriate amplitude.

1D *k*-space



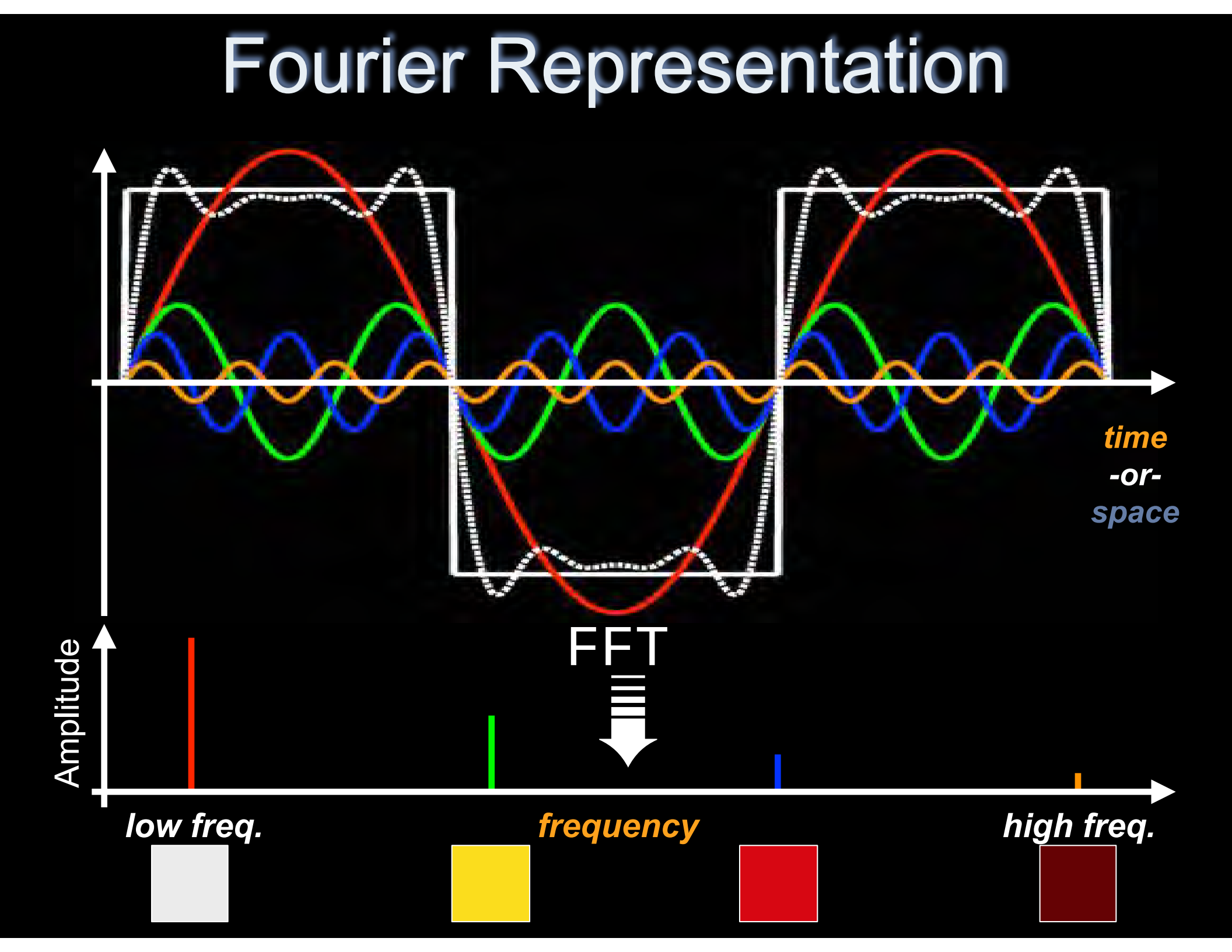
Any signal/image can be decomposed into a summation of sine waves of appropriate amplitude.

1D *k*-space



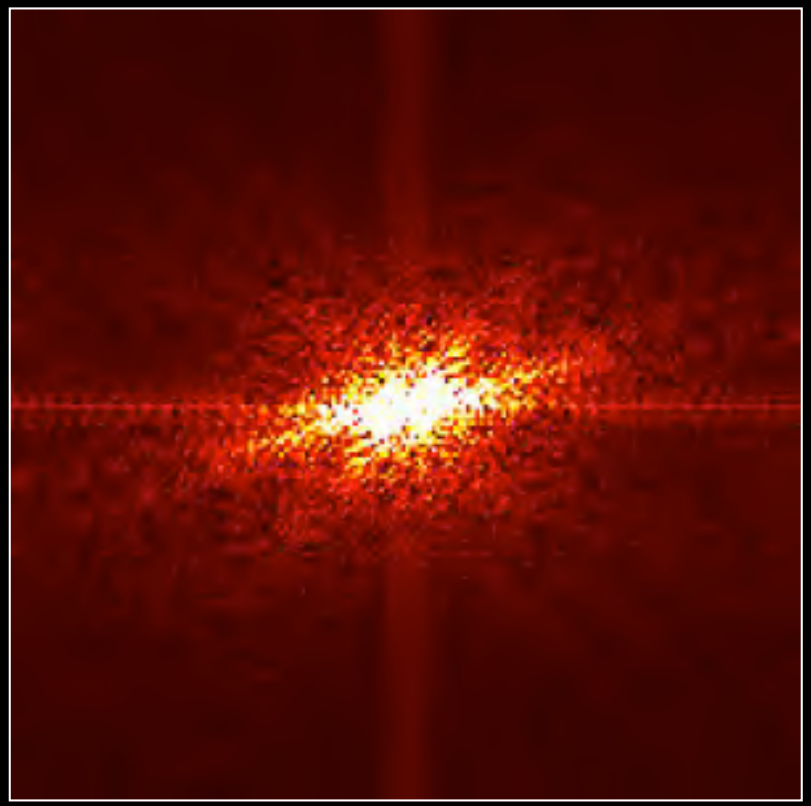
Any signal/image can be decomposed into a summation of sine waves of appropriate amplitude.

Fourier Representation



What is *k*-space?

k-space

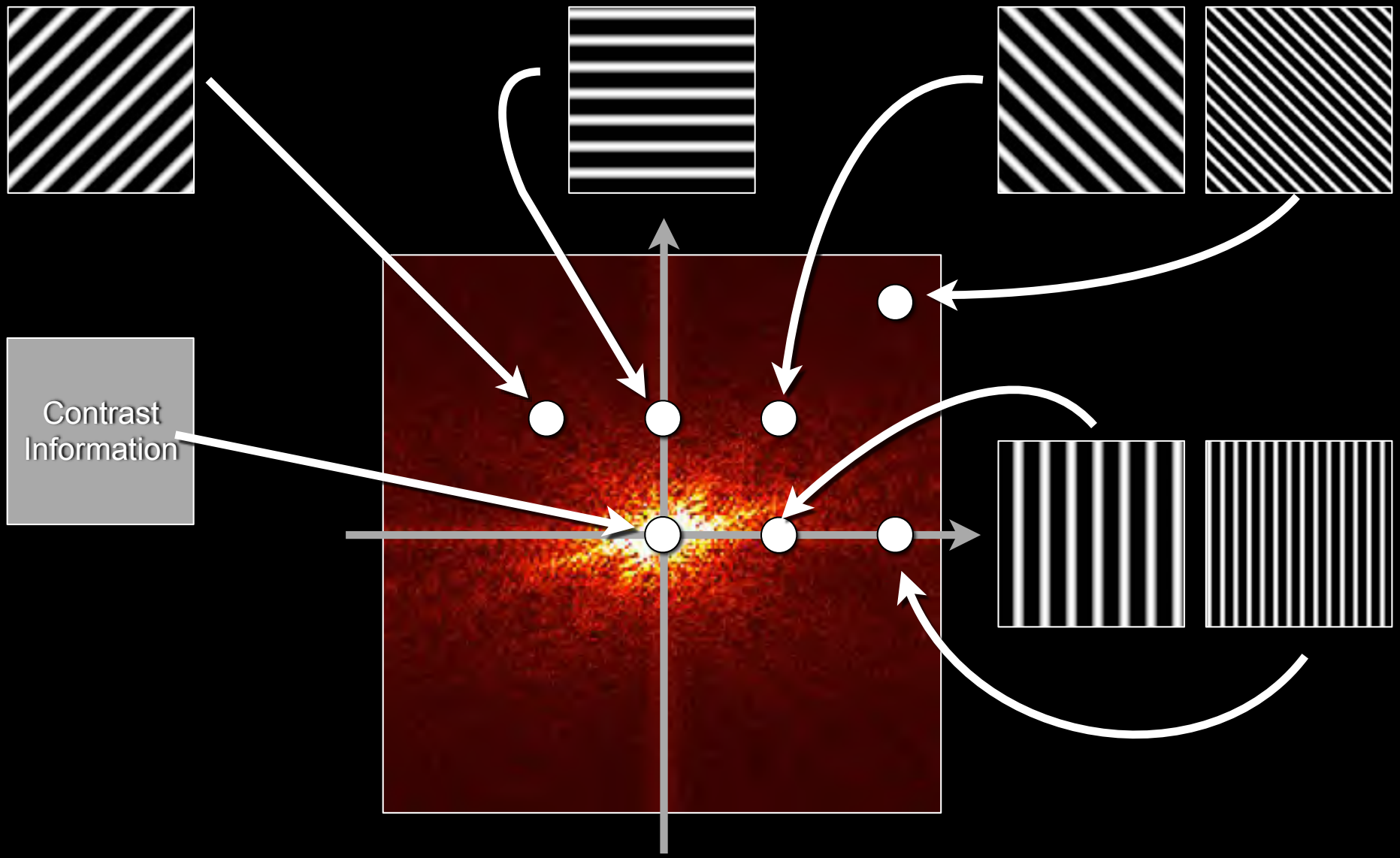


FFT

image space

***k*-space is the raw data collected by the scanner.**

What is *k*-space?

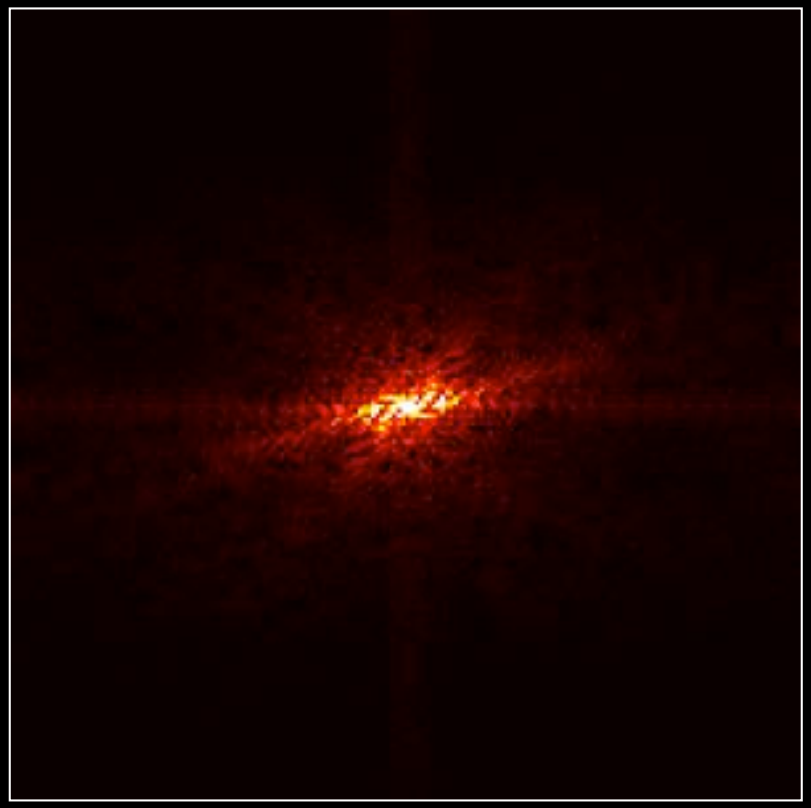


Points in *k*-space represent different patterns in an image.

k-space spikes

k-space

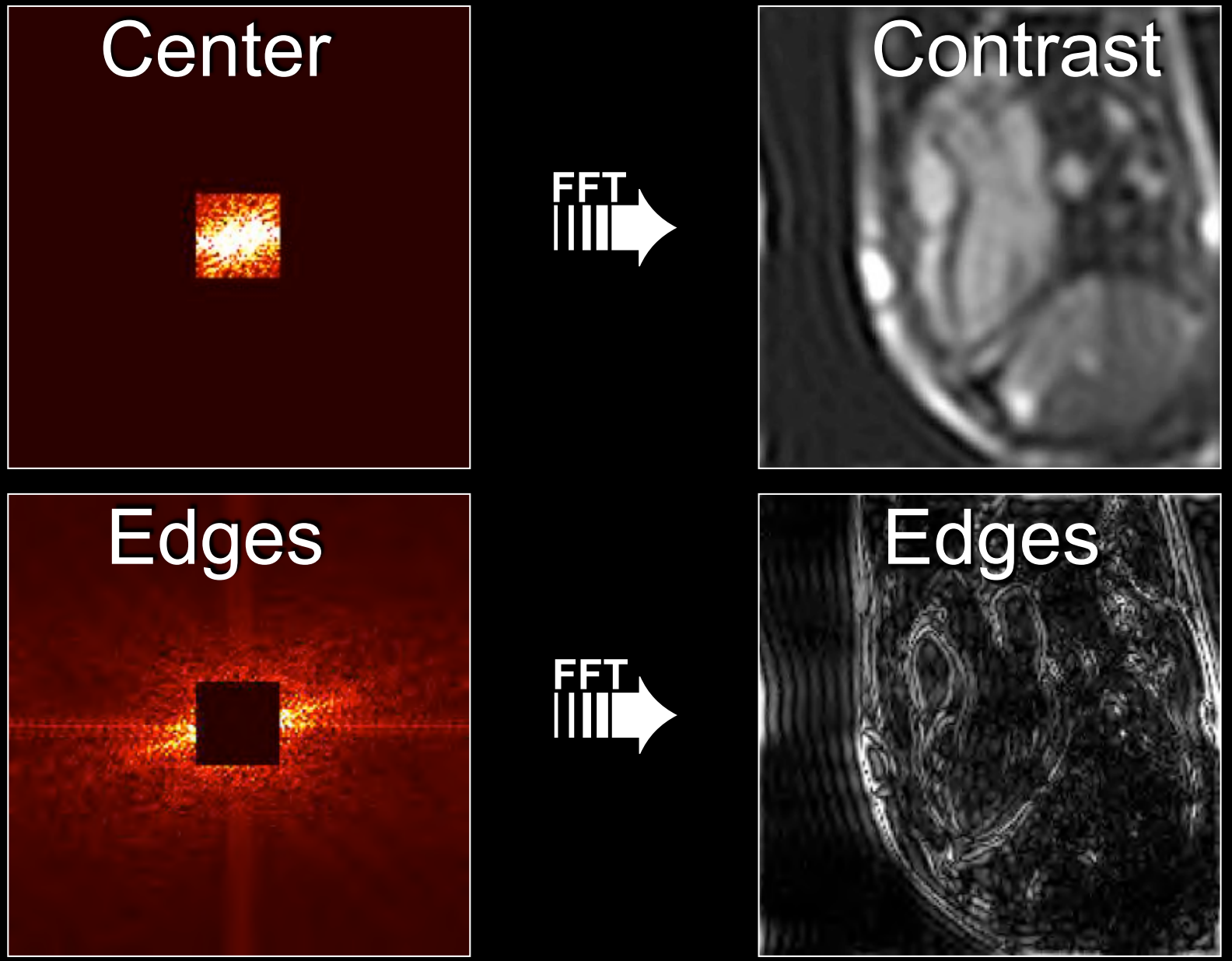
image space



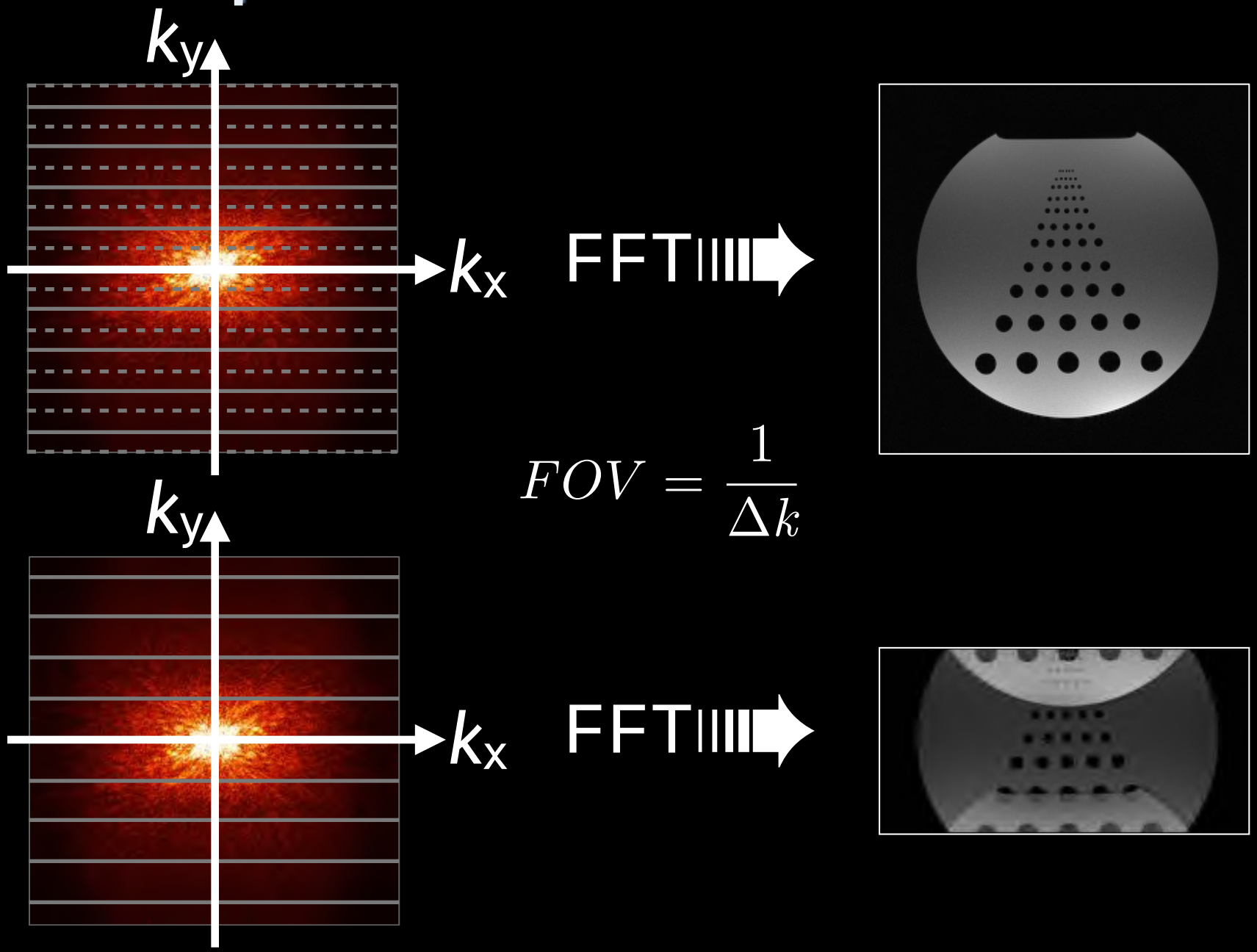
FFT

A *k*-space spike creates a banding artifact.

What is *k*-space?



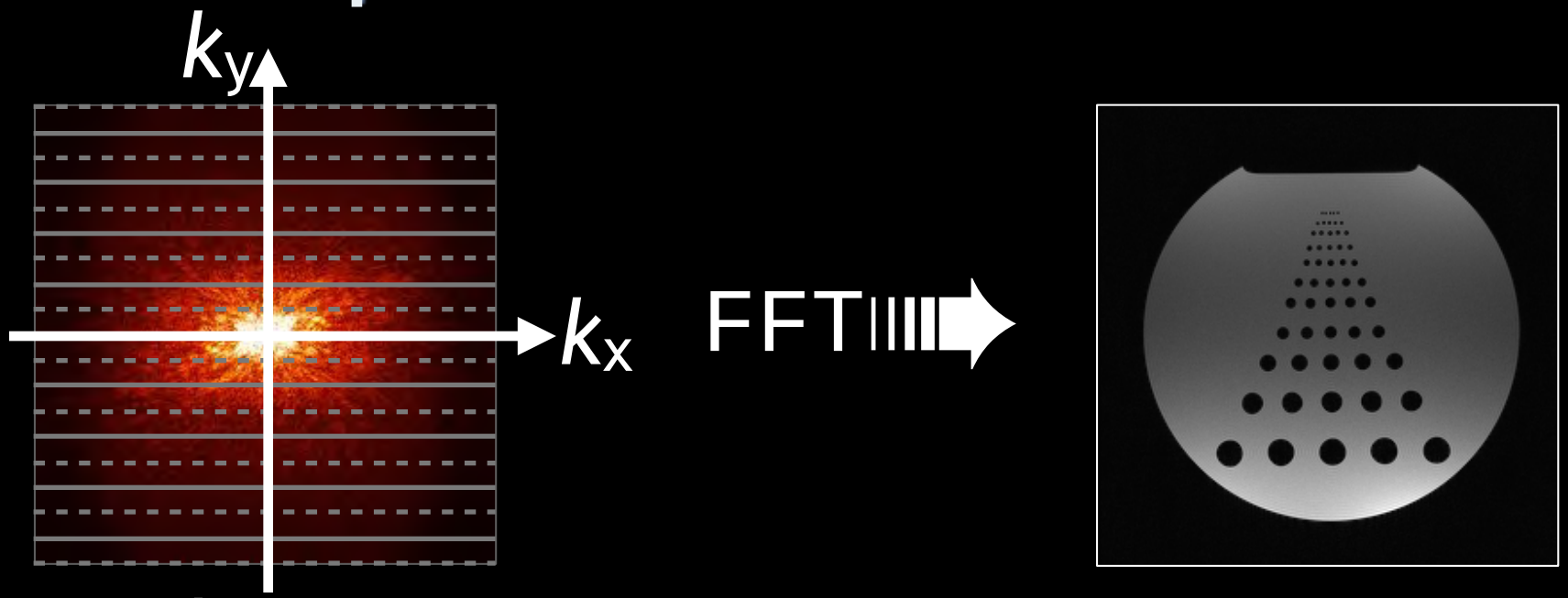
k-space and Field of View

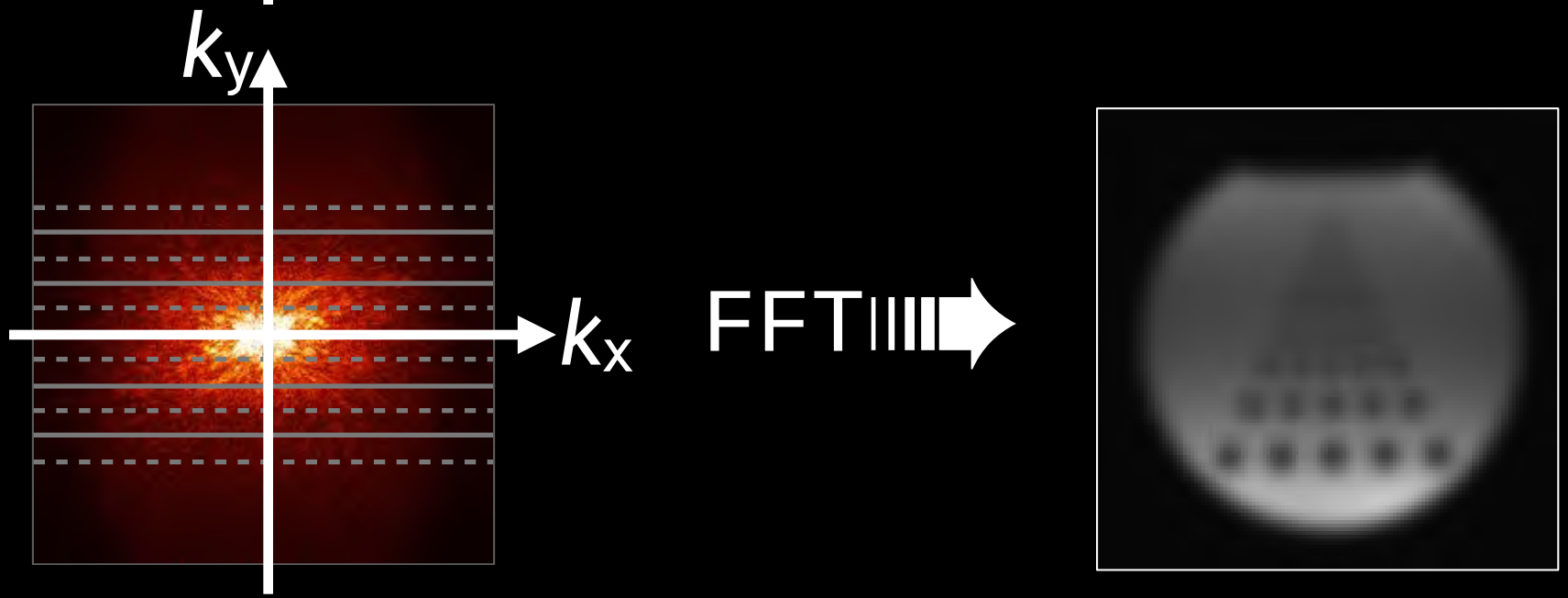


$$FOV = \frac{1}{\Delta k}$$

Uniformly skipping lines in *k*-space causes ***aliasing***.

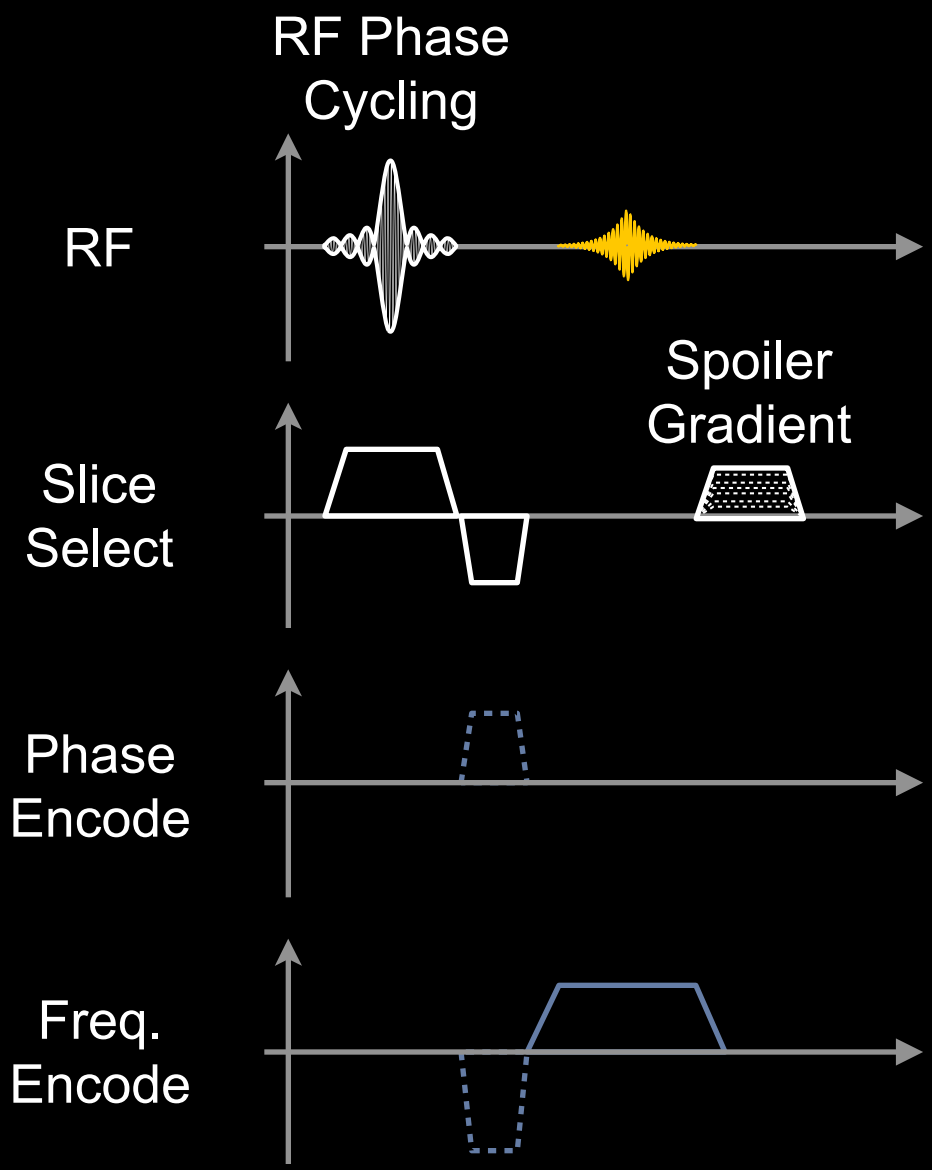
k-space and Resolution

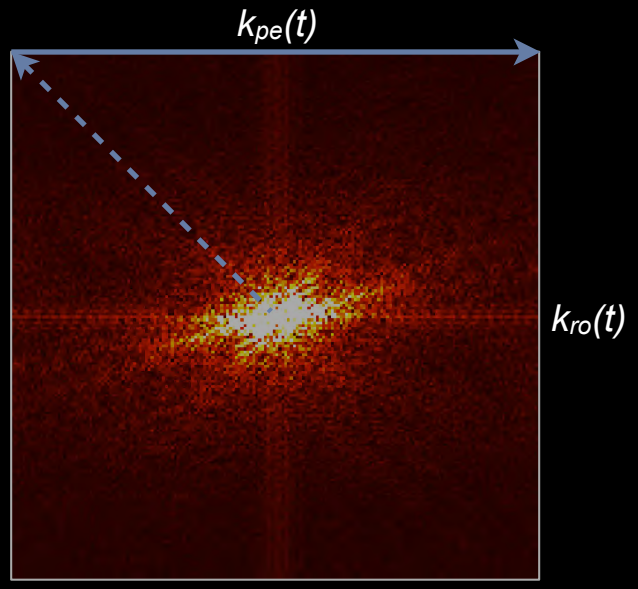




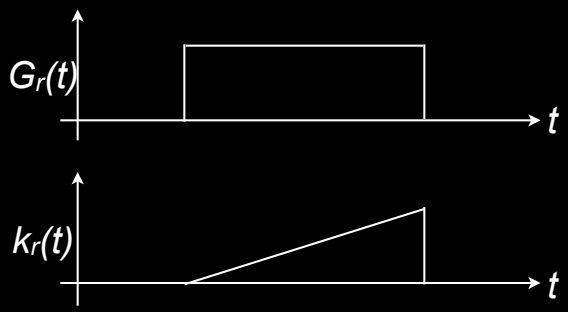
Acquiring fewer high phase encodes decreases resolution.

Where am I in *k*-space?



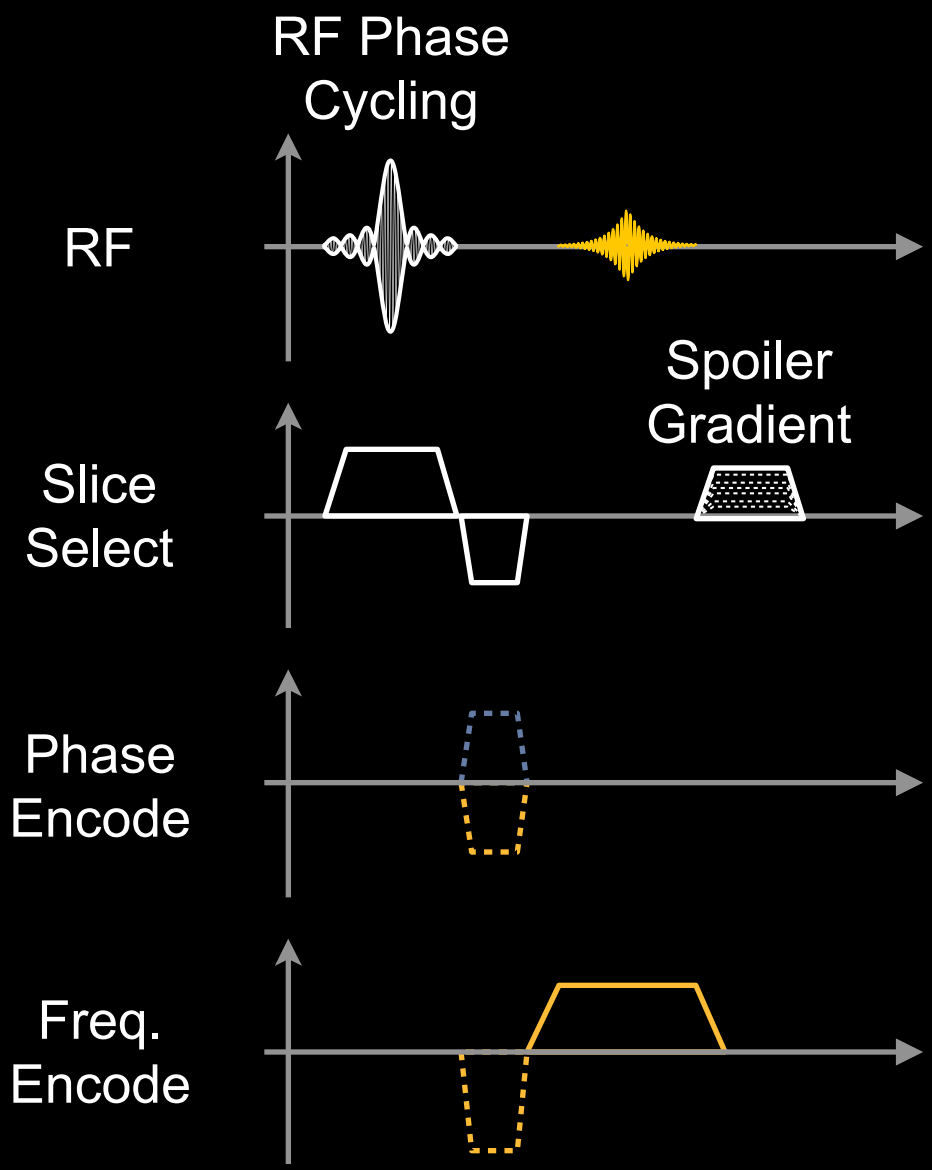


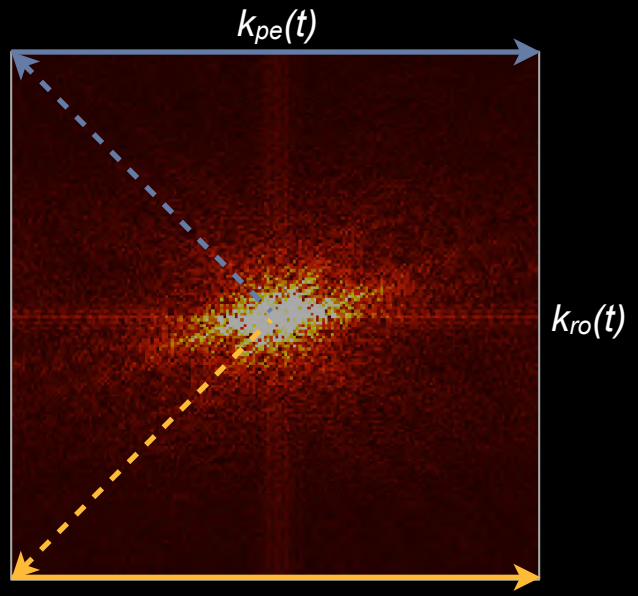
$$k_r(t) = \frac{\gamma}{2\pi} \int_0^{\tau} G_r(t) d\tau$$



Gradients move us around *k*-space.

Where am I in *k*-space?

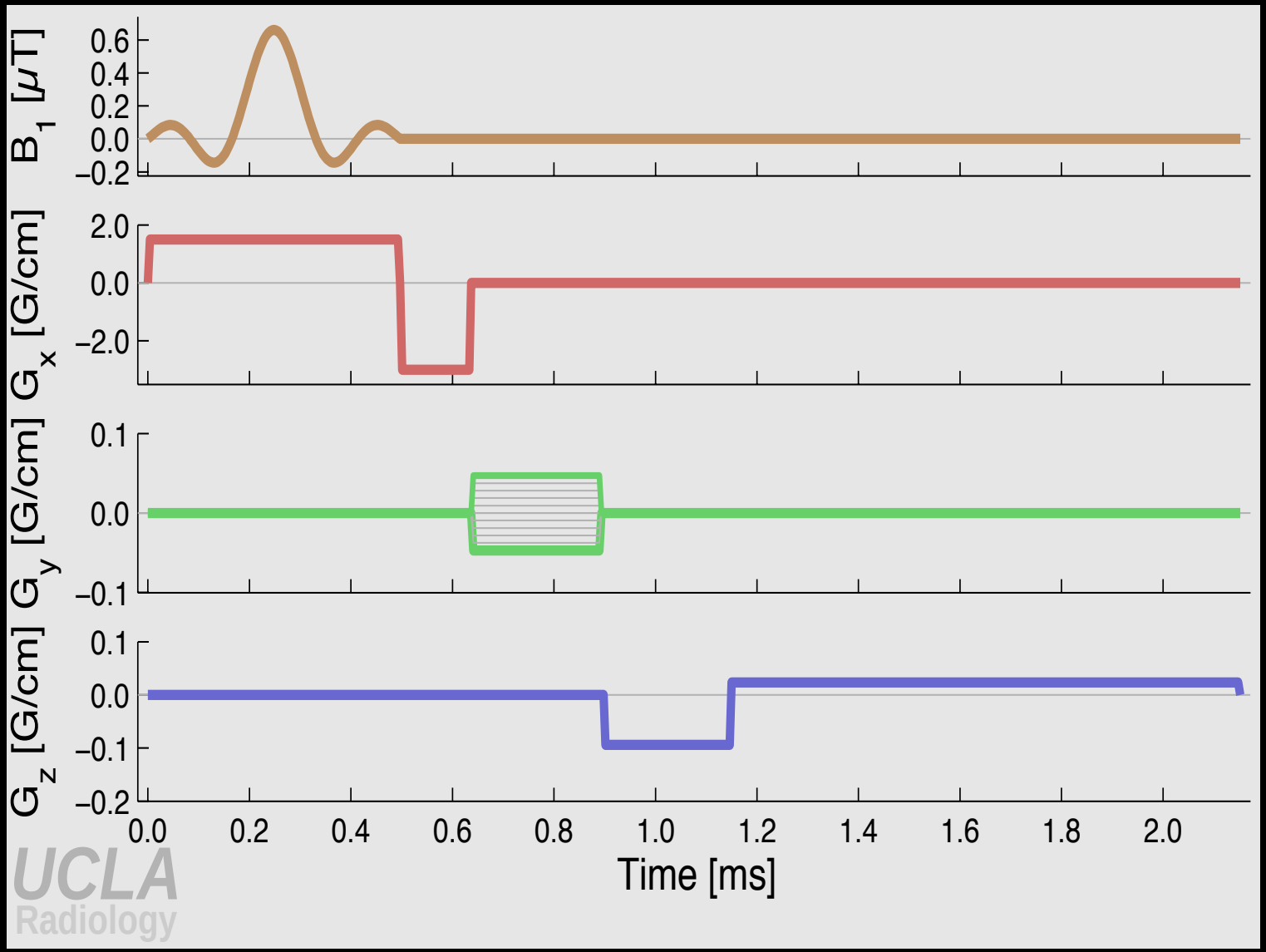




Gradients move us around *k*-space.

Image Formation

Spatial Localization



Slice Selection

Phase Encoding

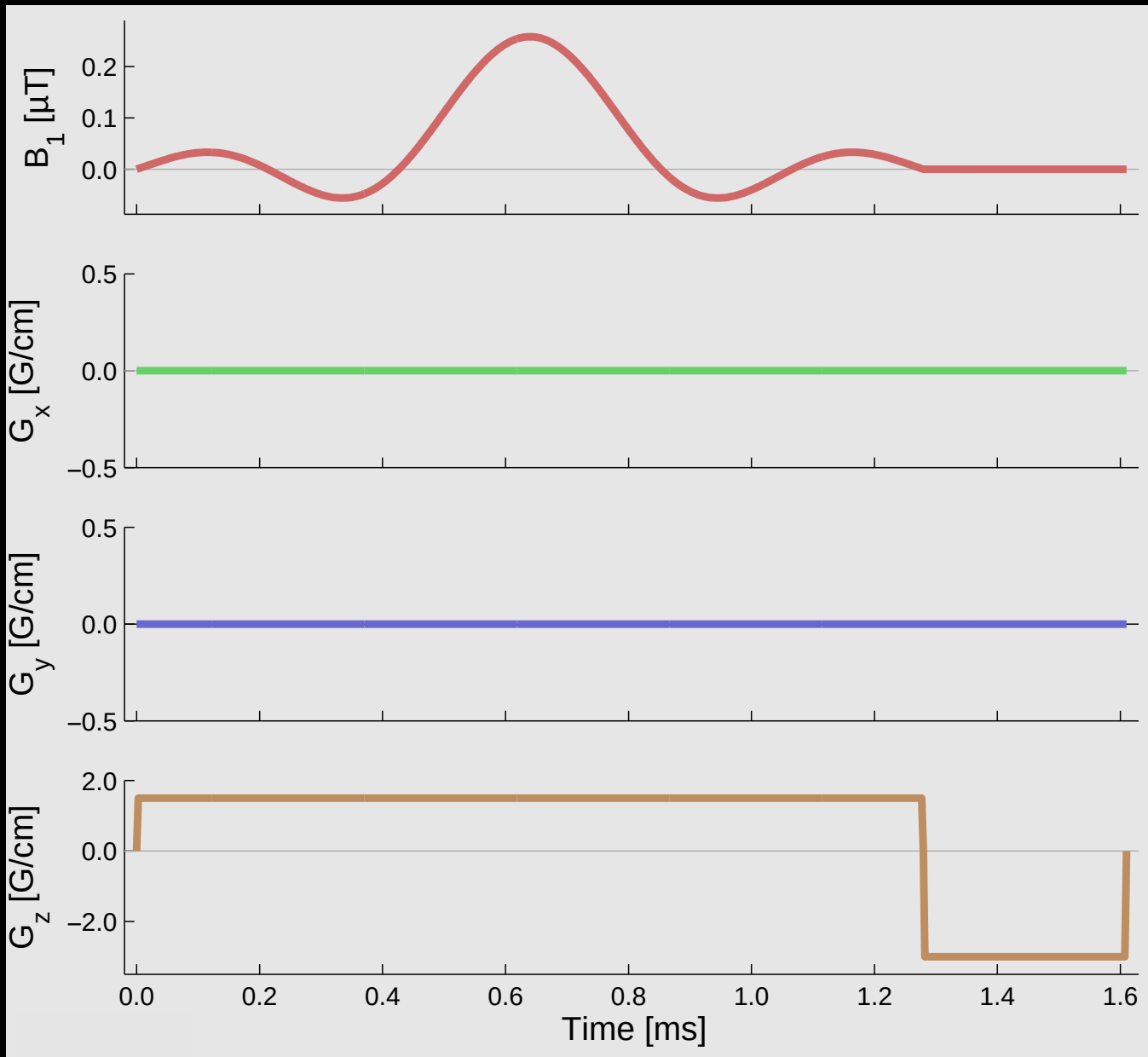
Frequency Encoding

Pulse Sequence Diagram - Timing diagram of the RF and gradient events that comprise an MRI pulse sequence.

Slice Selection

- **Consists of:**
  - **Slice selection gradient**
    - **Constant magnitude**
  - **RF (B_1) Pulse**
    - **Contains frequencies matched to slice of interest**
  - **Slice re-phasing gradient**
    - **Increases SNR**
    - **Re-phases spins within slice**
    - **AKA "slice refocusing gradient"**

Slice Selection

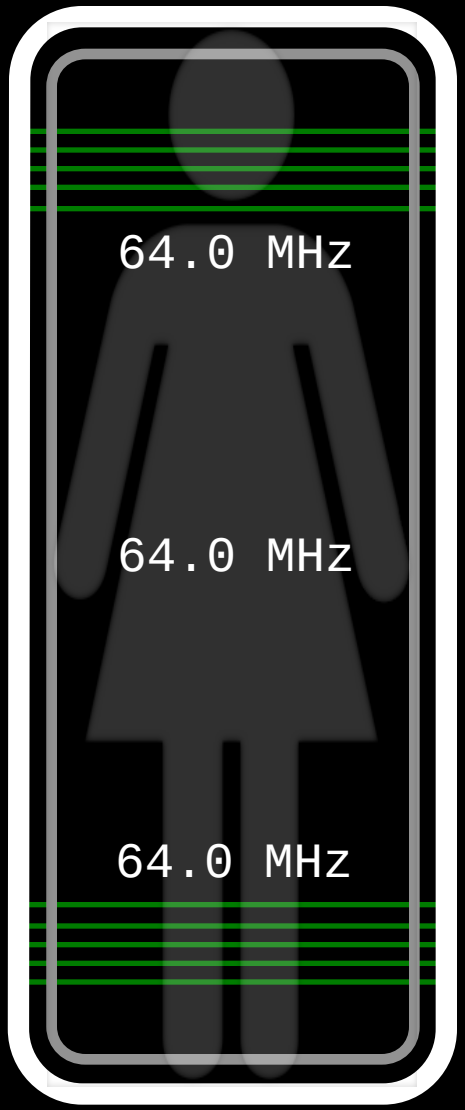


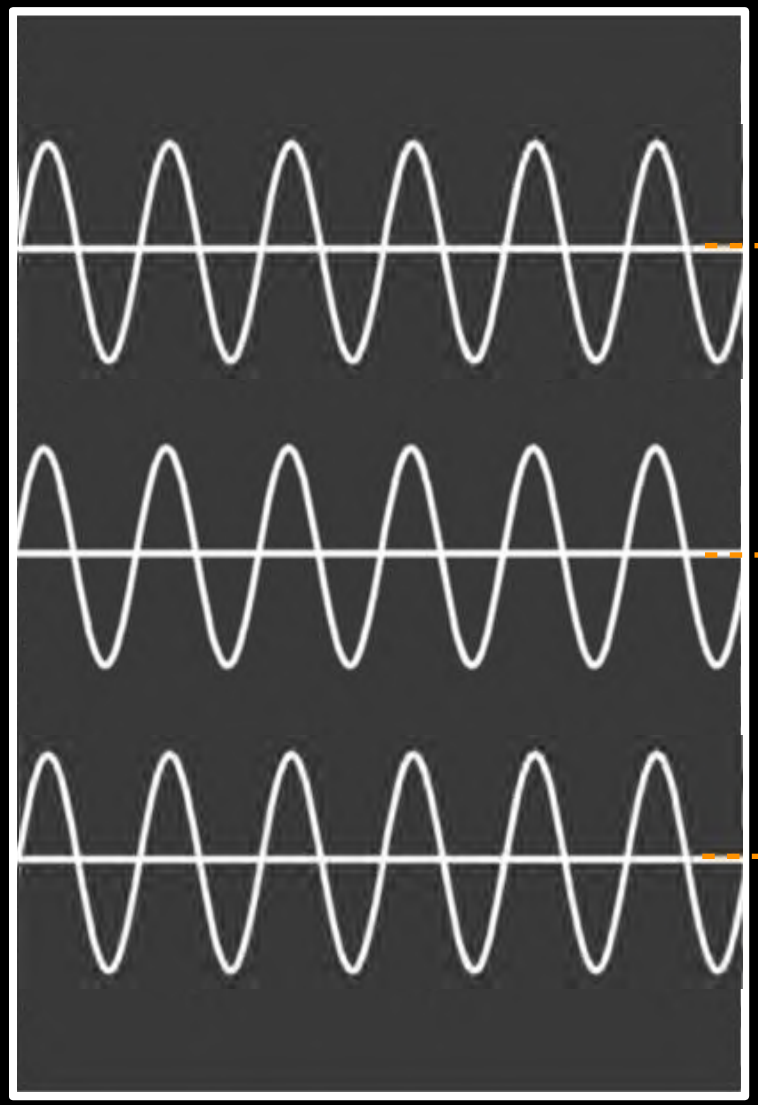
Tuned to frequencies (slice) of interest.

Creates a range of frequencies.

Re-phases spins in slice to increase signal-to-noise.

The Piano Analogy

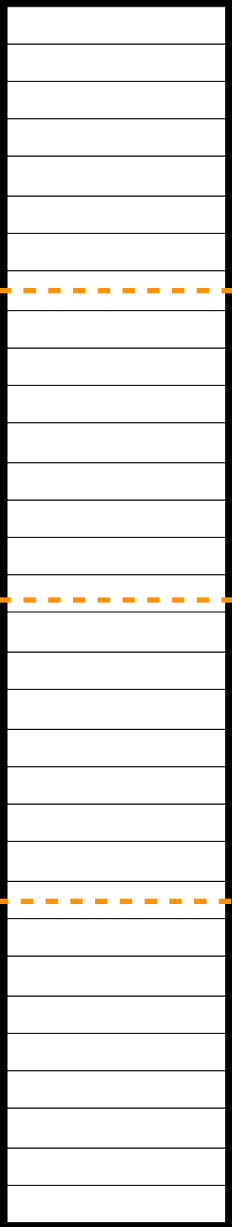




C4 261 Hz

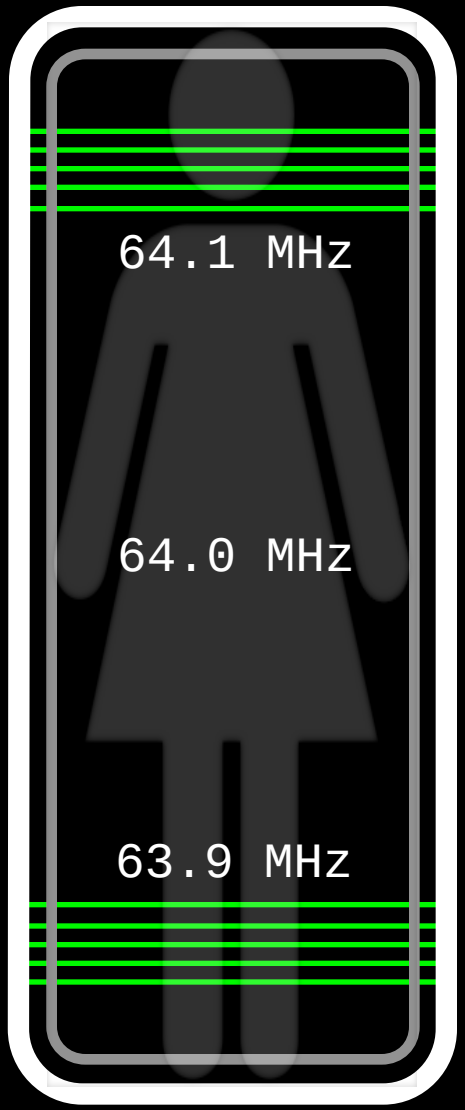
C4 261 Hz

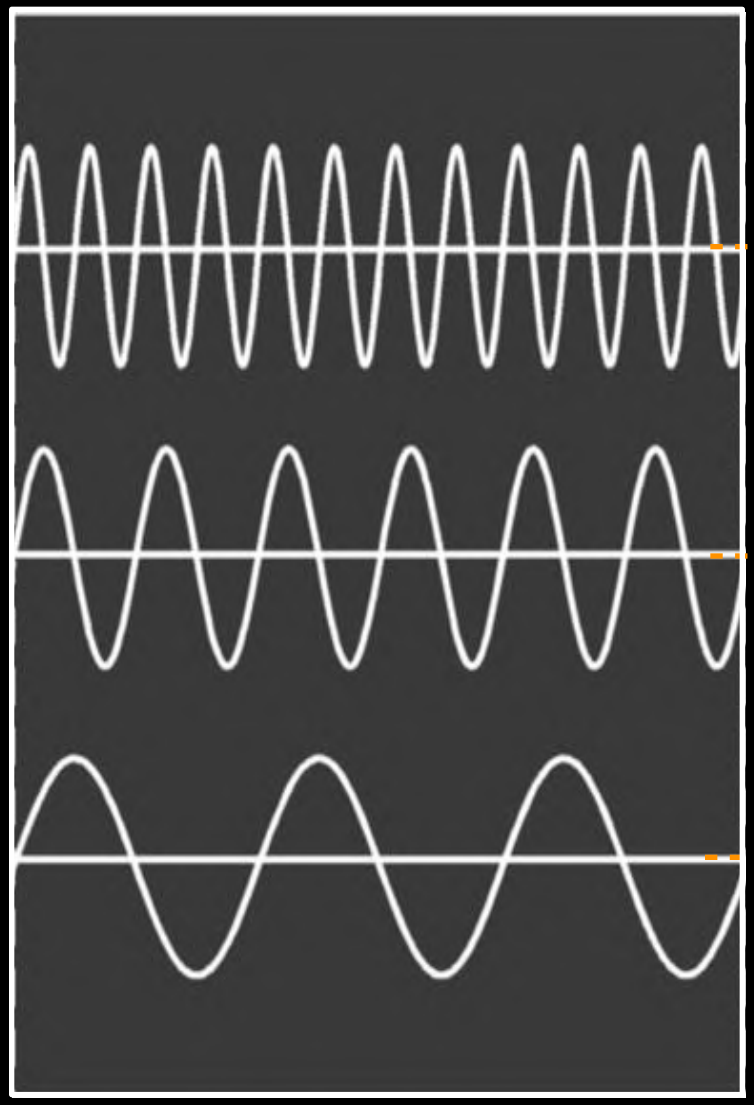
C4 261 Hz

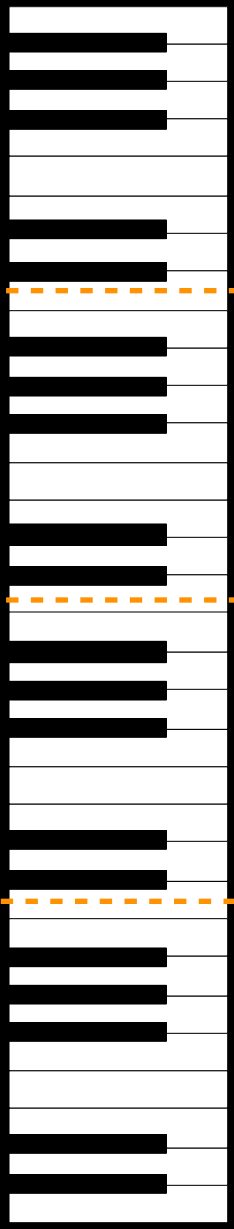


"Piano" Keyboard

The Piano Analogy







C5 523 Hz

C4 261 Hz

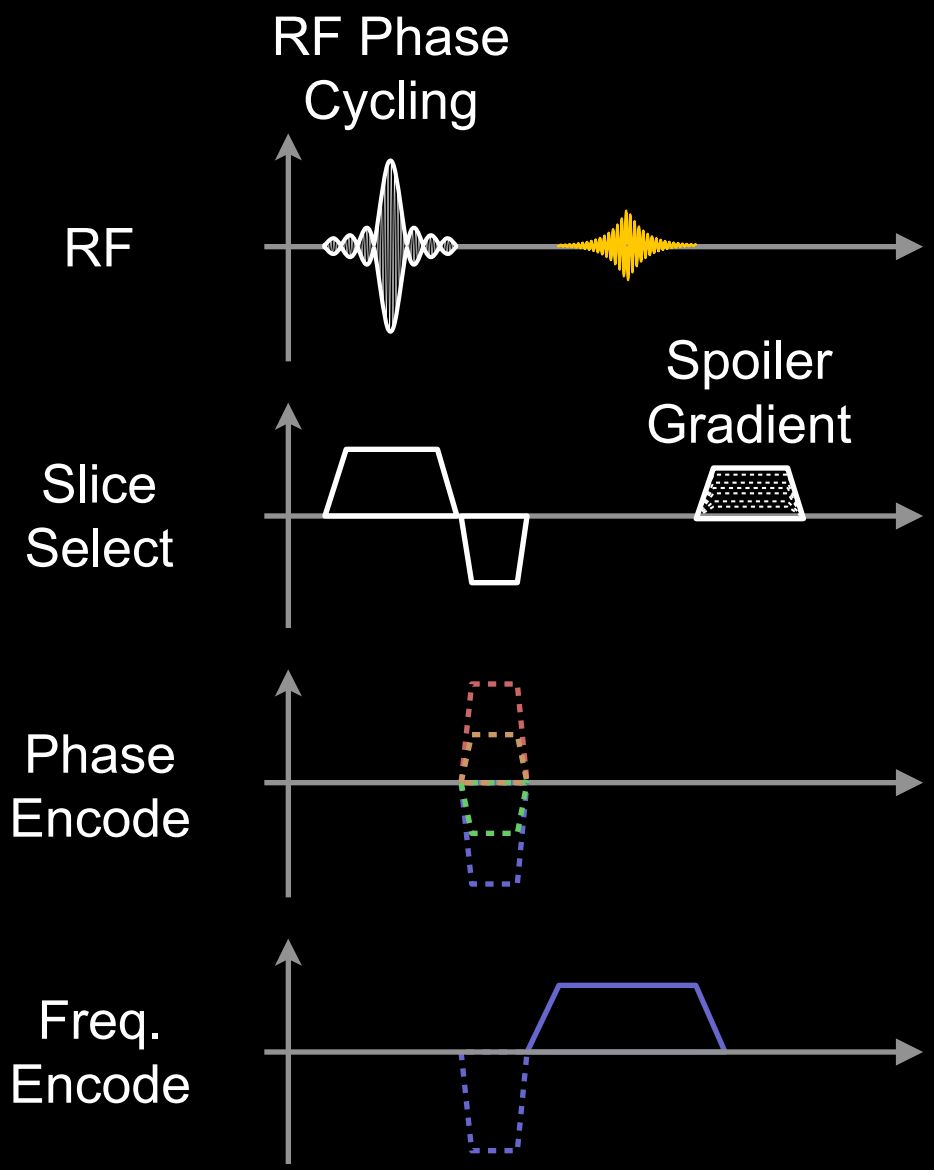
C3 130 Hz

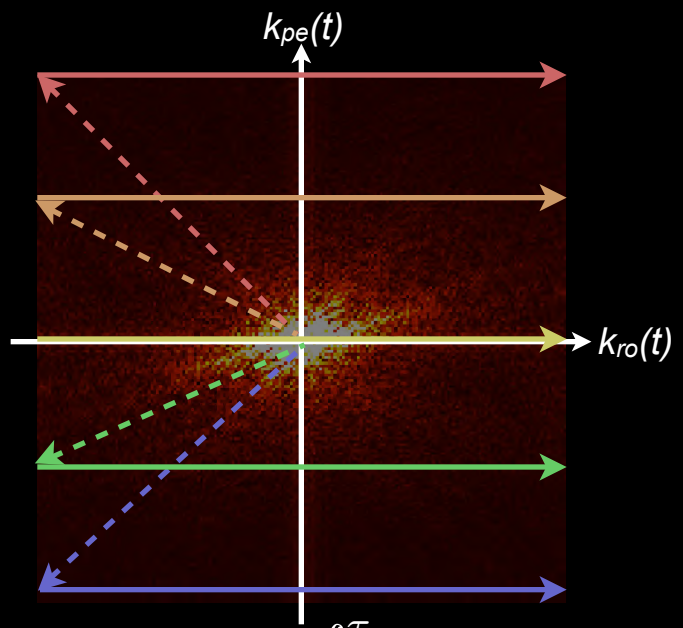
Piano Keyboard

Phase Encoding

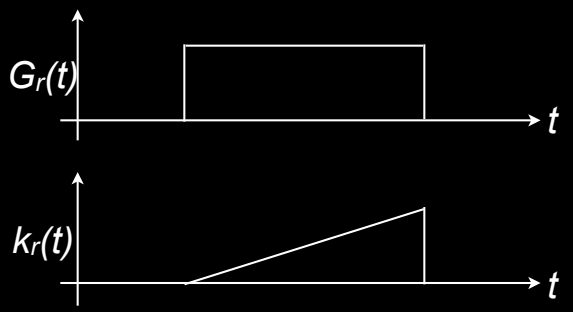
- Consists of:
  - Phase encoding gradient
    - Magnitude changes with each TR
    - Can be played with other gradients
      - Crushers, Slice-selection rephaser, readout dephasing
- Used with Cartesian imaging
- After excitation, before readout
- Adds linear spatial variation of phase
- Phase encode in
  - one direction for 2D imaging
  - two directions for 3D imaging
- Only one PE step per echo

Where am I in *k*-space?





$$k_r(t) = \frac{\gamma}{2\pi} \int_0^{\tau} G_r(t) d\tau$$

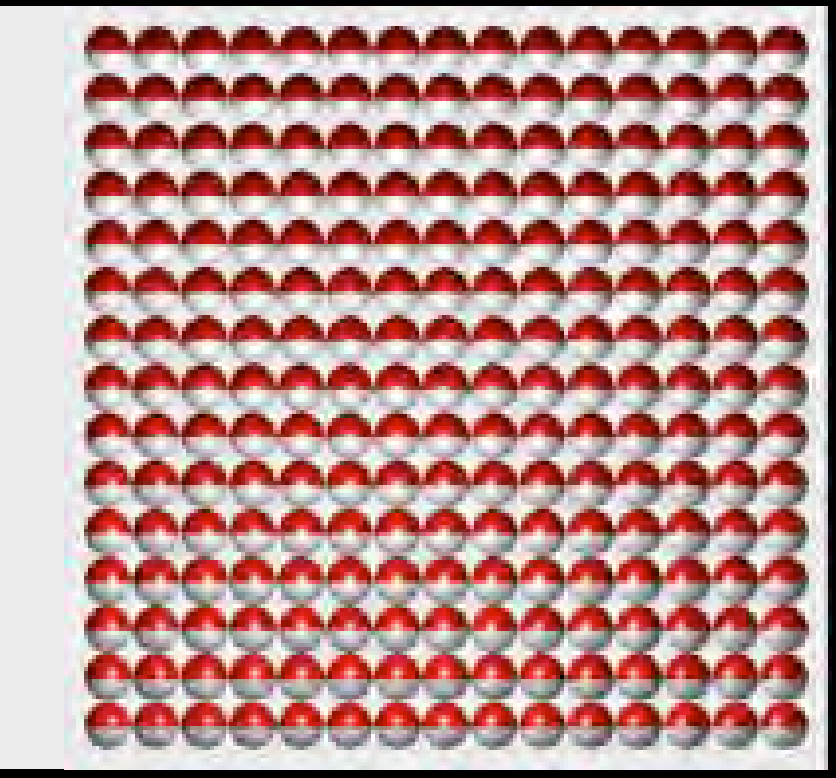


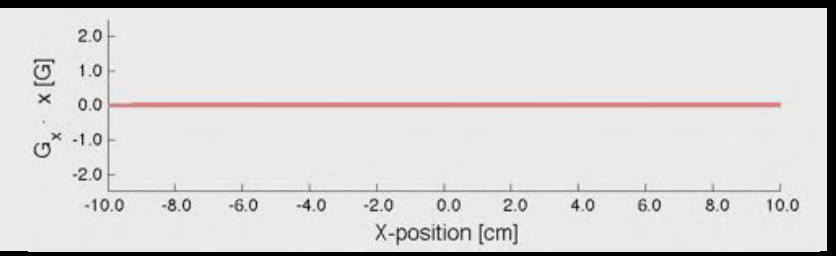
Frequency Encoding

- **Consists of:**
  - **Frequency encoding gradient**
    - **Constant magnitude for Cartesian imaging**
  - **No simultaneous**
    - **RF (B_1)**
    - **Other gradients**
      - **phase encoding, slice encoding, crushers**
  - **Readout pre-phasing gradient**
    - **Prepares spin phase so peak echo amplitude occurs at middle of readout (TE)**
    - **AKA "readout de-phasing gradient"**
- **Adds linear spatial variation of frequency**
- **Helps form an *echo***

Frequency Encoding

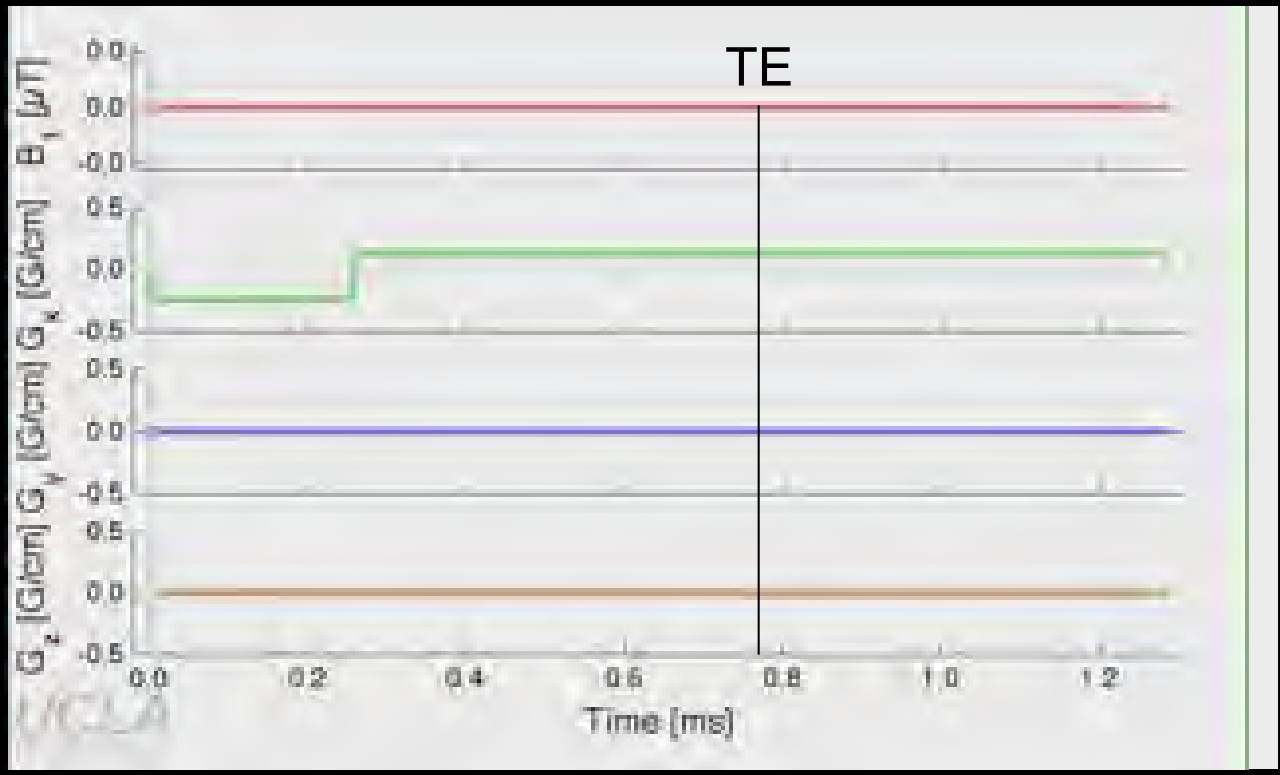
B_0-G_x•x B_0 B_0+G_x•x





Applied Magnetic Field

Frequency Encoding



Dipoles to Images

$\vec{\mu}$	Magnetic Moment	B_0
$\downarrow$		
$\vec{M}_z$	Bulk Magnetization	B_1
$\downarrow$		
$\vec{M}_{xy}$	Transverse Magnetization	Coil
$\downarrow$		
$S(t)$	Received Signal	Gradients
$\downarrow$		
$S\left(\vec{k}\right)$	k-space signal	FFT
$\downarrow$		
$I(\vec{x})$	Image	

Introduction to MRI

Part 1: Image Formation
Part 2: Image Contrast

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Magnetic Resonance Research Labs

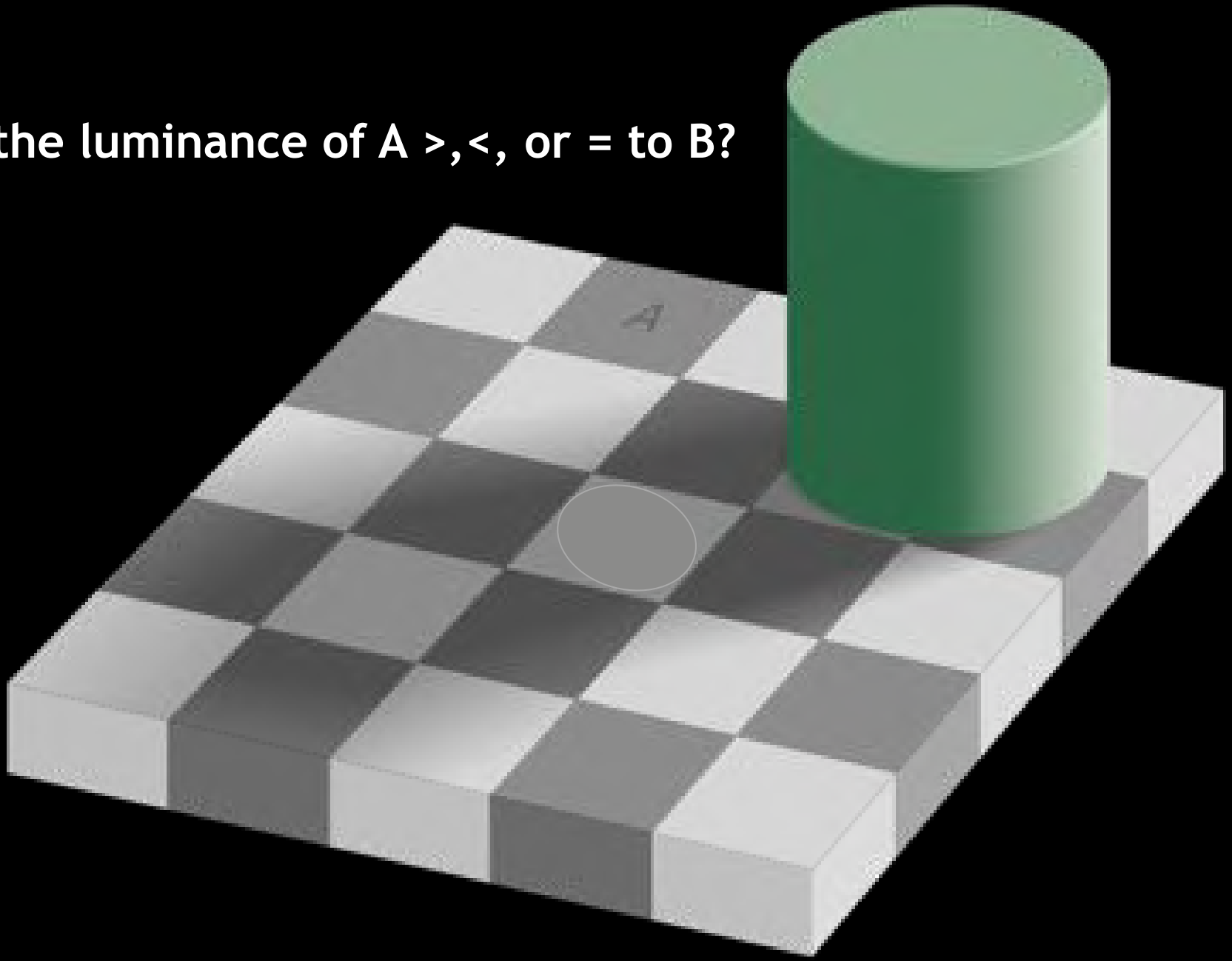
David Geffen School of Medicine

UCLA Radiology

Relaxation & Image Contrast

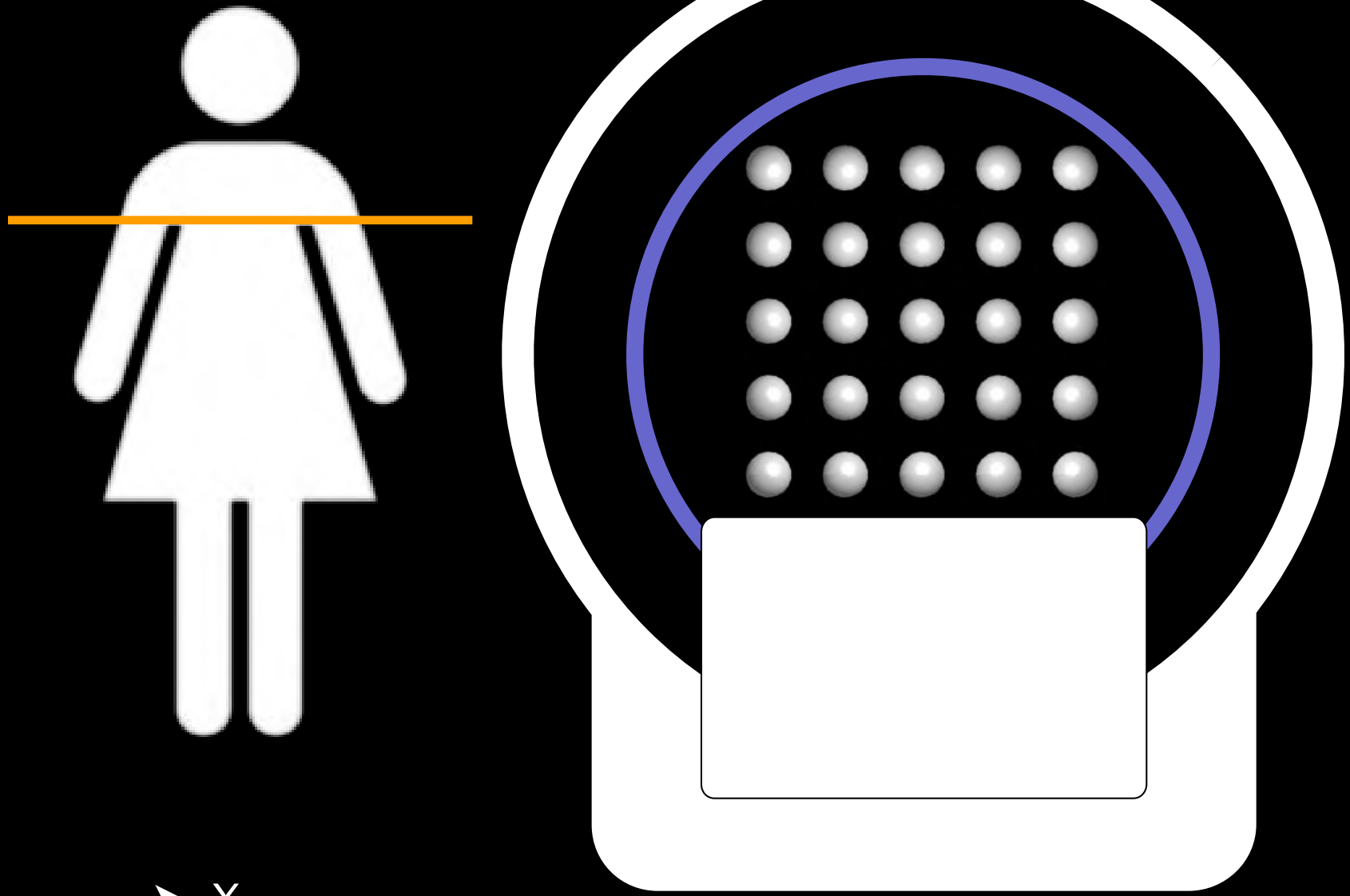
Why Image Contrast?

Is the luminance of A >,<, or = to B?



Humans perceive contrast better and luminance.

Excitation and Relaxation



The magnetization relaxes after excitation.

1952 Nobel Prize in Physics

“for their development of new methods for nuclear magnetic precision measurements and discoveries in connection therewith“

Felix Bloch
b. 23 Oct 1905
d. 10 Sep 1983

Edward Purcell
b. 30 Sep 1912
d. 07 Mar 1997

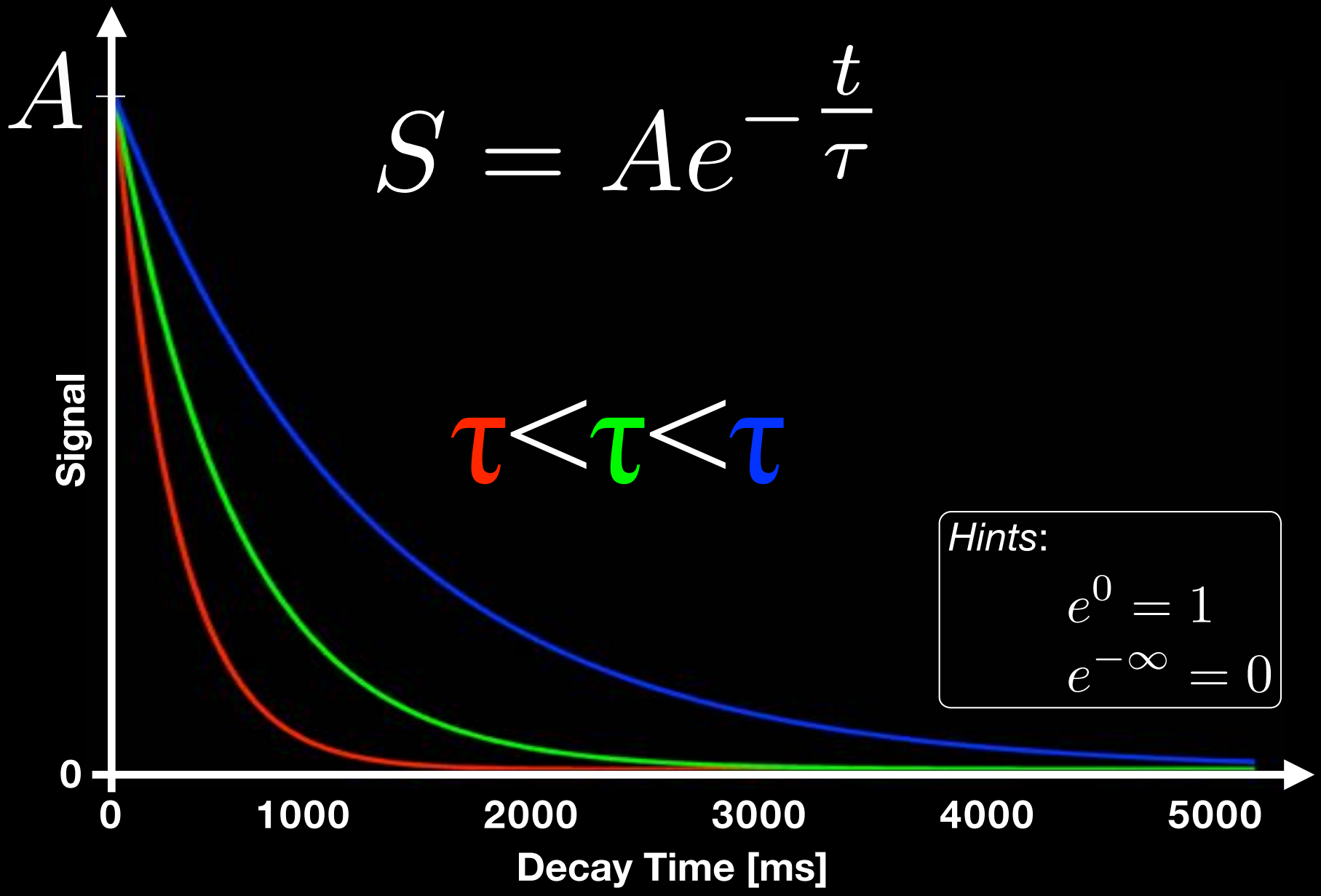
Bloch Equations

$$\frac{d\vec{\mathrm{M}}}{dt} = \underbrace{\vec{\mathrm{M}} \times \gamma\vec{\mathrm{B}}}_{\text{Precession}} - \underbrace{\frac{M_x\hat{\mathrm{i}} + M_y\hat{\mathrm{j}}}{T_2}}_{\substack{\text{Transverse} \\ \text{Relaxation } (M_{xy})}} - \underbrace{\frac{(M_z - M_0)\,\hat{\mathrm{k}}}{T_1}}_{\substack{\text{Longitudinal} \\ \text{Relaxation } (M_z)}} + \underbrace{\mathbf{D}\nabla^2\vec{\mathrm{M}}}_{\text{Diffusion}}$$

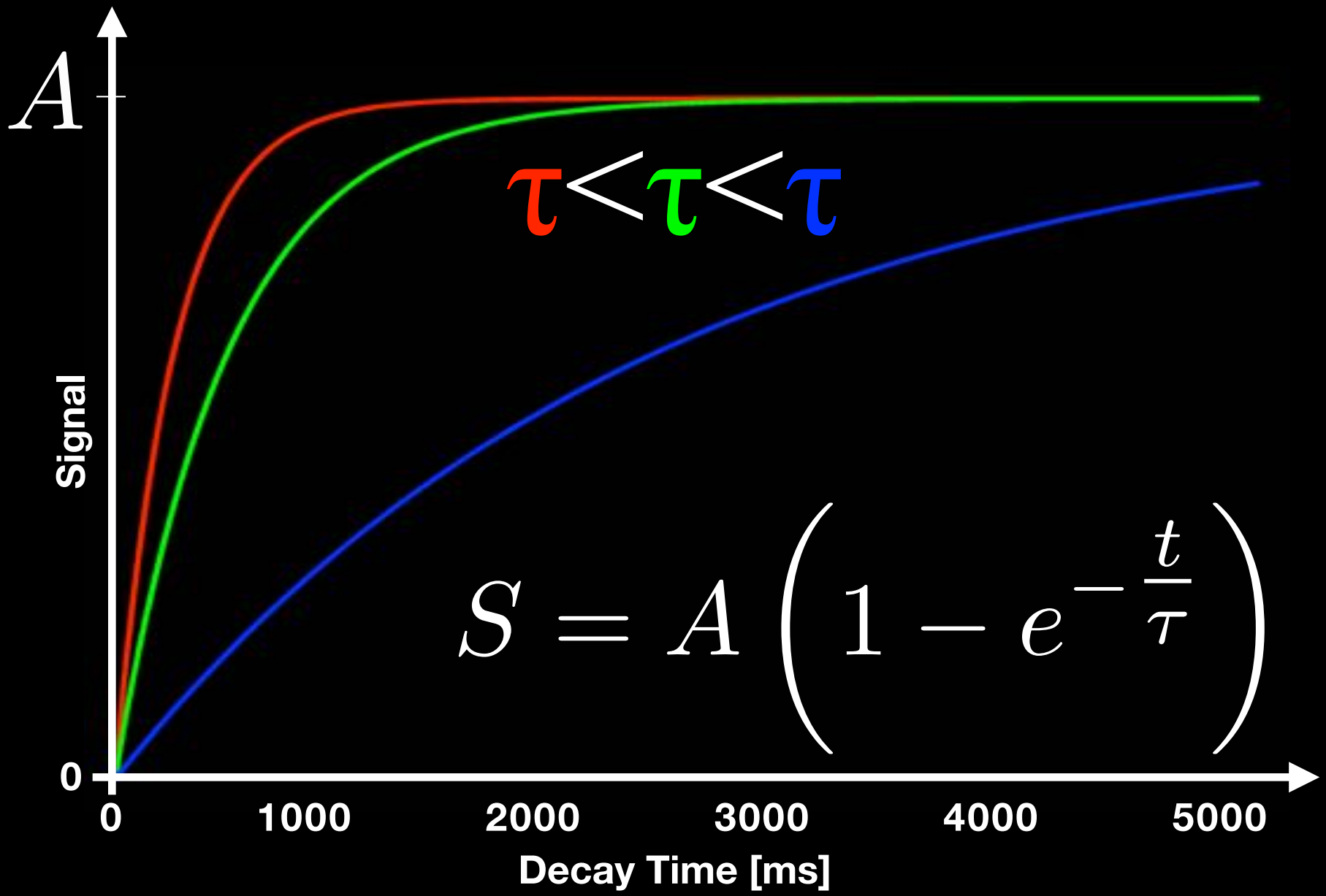
- Precession
  - Magnitude of $\vec{\mathrm{M}}$ unchanged
  - Phase (rotation) of $\vec{\mathrm{M}}$ changes due to $\vec{\mathrm{B}}$
- Relaxation
  - T_1 changes are slow O(100ms)
  - T_2 changes are fast O(10ms)
  - Magnitude of M can be ZERO
- Diffusion
  - Spins are thermodynamically driven to exchange positions.

Bulk magnetization has two key components – longitudinal and transverse.

Exponential Functions - Decay



Exponential Functions - Recovery



Longitudinal & Transverse Relaxation

$$M_z(t) = \underbrace{M_z^0 e^{-\frac{t}{T_1}}}_{\text{Initial Condition}} + \underbrace{M_0\left(1 - e^{-\frac{t}{T_1}}\right)}_{\text{Return to Equilibrium}}$$

$$M_{xy}(t) = \underbrace{M_{xy}^0 e^{-t/T_2}}_{\substack{\text{Initial Condition} \\ \text{Return to Equilibrium}}}$$

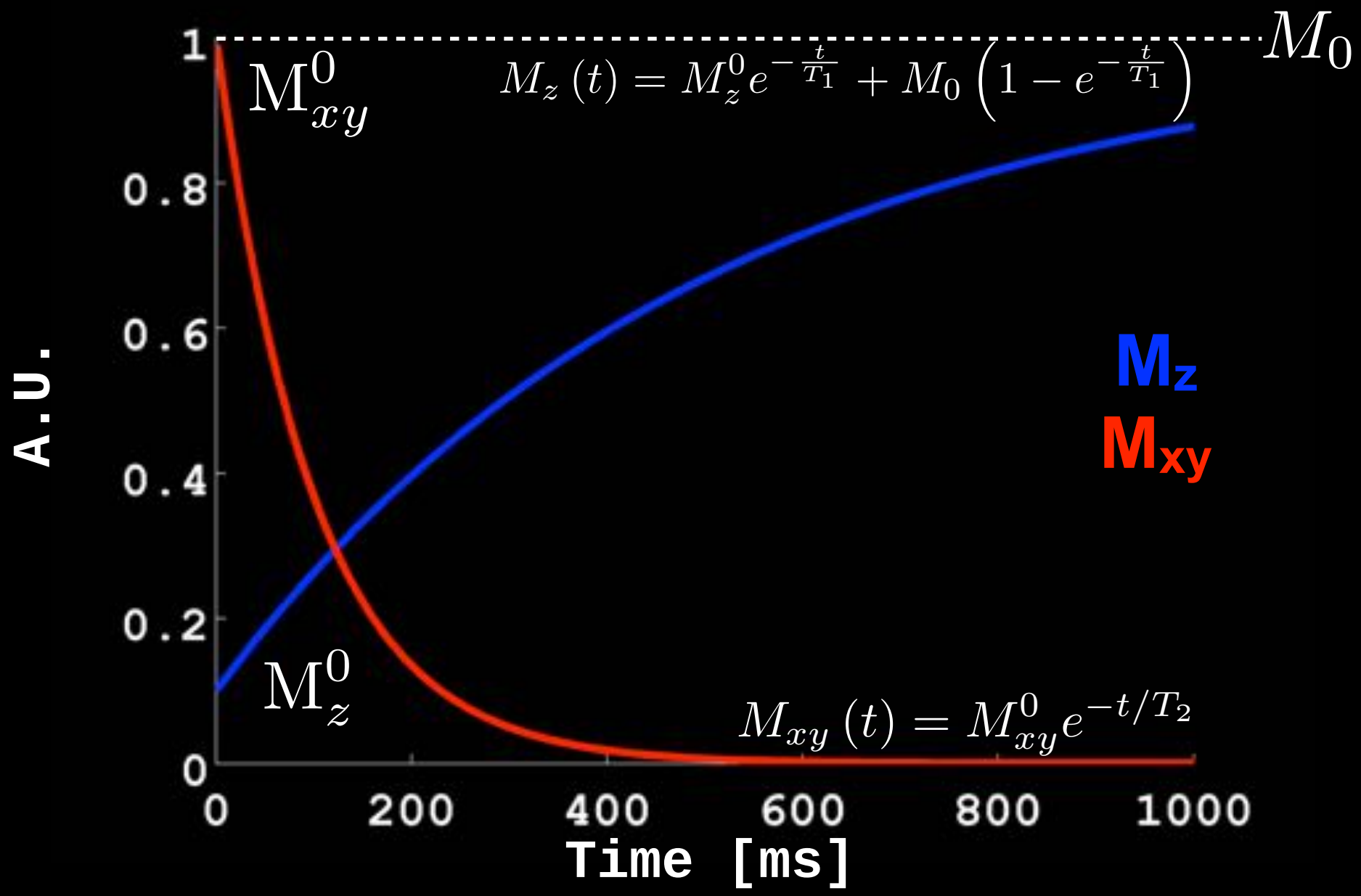
General solutions to the Bloch equations with relaxation in the rotating frame during free precession.

T_1 and T_2 Values @ 1.5T

Tissue	T_1 [ms]	T_2 [ms]
gray matter	925	100
white matter	790	92
muscle	875	47
fat	260	85
kidney	650	58
liver	500	43
CSF	2400	180

Soft tissues have specific relaxation properties.

T_1 & T_2 Relaxation



M_z returns to equilibrium. M_{xy} decays to zero.

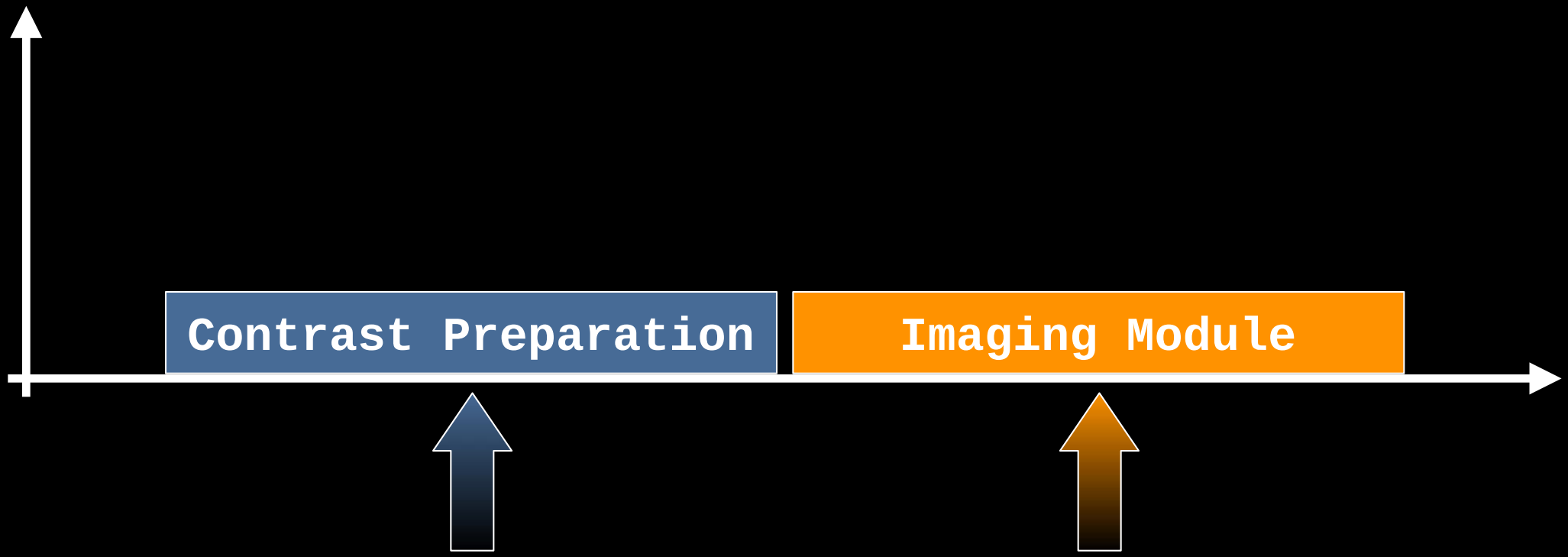
T_1 Relaxation

- Longitudinal or spin-lattice relaxation
- Typically, (10s ms)<T_1< (100s ms)
- T_1 is long for
  - Small molecules (water)
  - Large molecules (proteins)
- T_1 is short for
  - Fats and intermediate-sized molecules
- T_1 increases with increasing B_0
- T_1 decreases with contrast agents
- Short T_1s are bright on T_1-weighted image

T_2 Relaxation

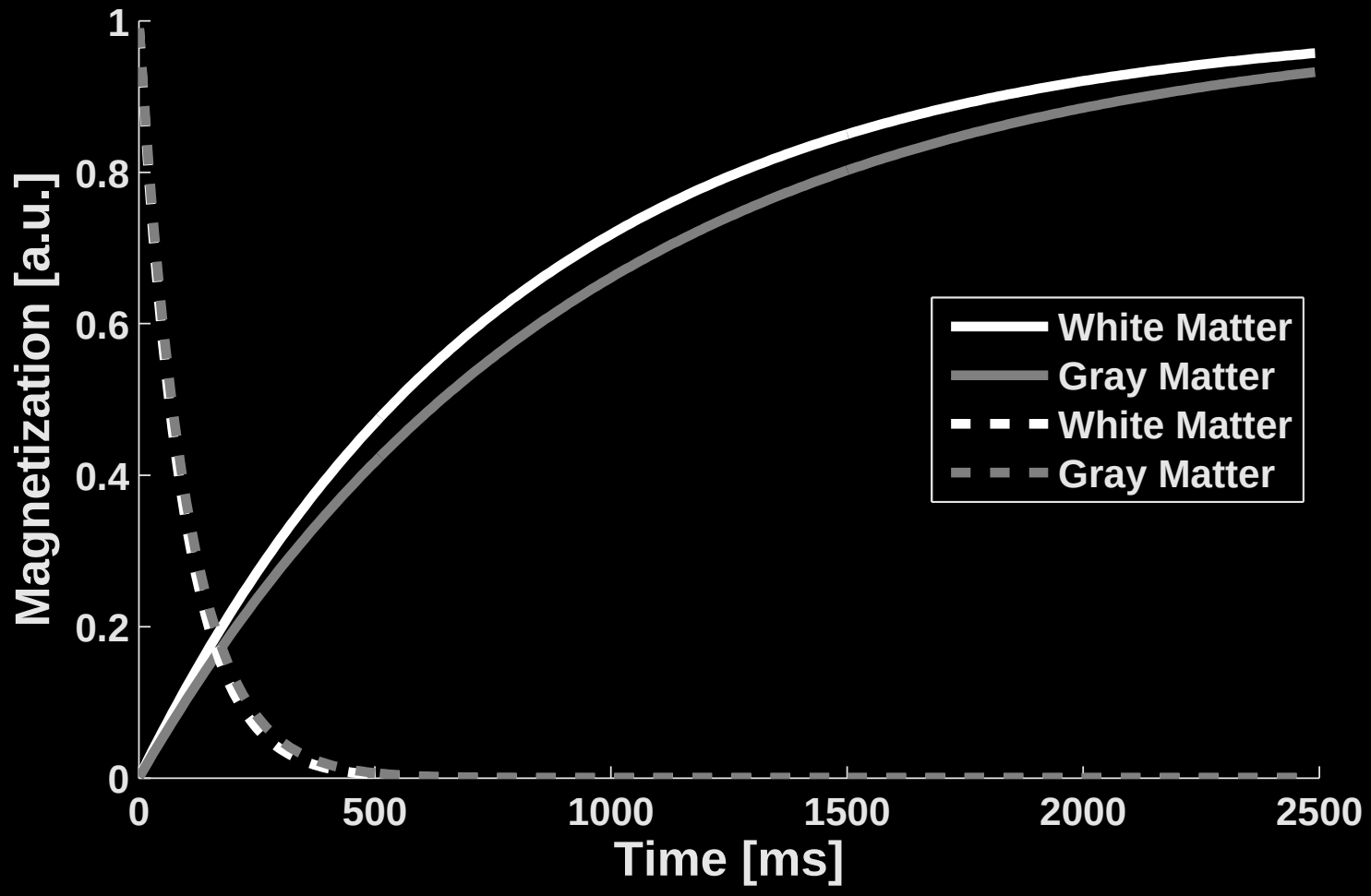
- Transverse or spin-spin relaxation
  - Molecular interaction causes spin dephasing
- Typically, T_2<(10s ms)
- Increasing molecular size, decrease T_2
  - Large molecules have a short T_2
  - Fat has a short T_2
- Increasing molecular mobility, increases T_2
  - Liquids have long T_2s
  - CSF, edema
- Increasing molecular interactions, decreases T_2
  - Solids have short T_2s
- T_2 relatively independent of B_0
- T_2 always < T_1
- Long T_2 is bright on T_2 weighted image

Pulse Sequences



T_1 & T_2 Relaxation

Tissue	T_1 [ms]	T_2 [ms]
gray matter	925	100
white matter	790	92



T_2^* Relaxation

- The “observed” transverse relaxation time constant
- Intravoxel Spin Dephasing
  - Spin-spin (T_2) dephasing combined with...
    - Irreversible
  - Off-resonance
    - B_0 inhomogeneity
    - Susceptibility differences (macro and micro)
    - Chemical shift effects
    - Reversible!!!
- Off-resonance can be rephased with a spin echo
  - Not with a gradient echo!
- $T_2^* < T_2$

T_2^* Relaxation

$$\frac{1}{T_2^*} = \frac{1}{T_2} + \gamma \delta B_0$$

Irreversible Losses

Reversible Losses

T_2^* Relaxation

$$\frac{1}{T_2^*} = \frac{1}{T_2} + \frac{1}{T_2'}$$

Irreversible Losses

Reversible Losses

T_2^* Relaxation

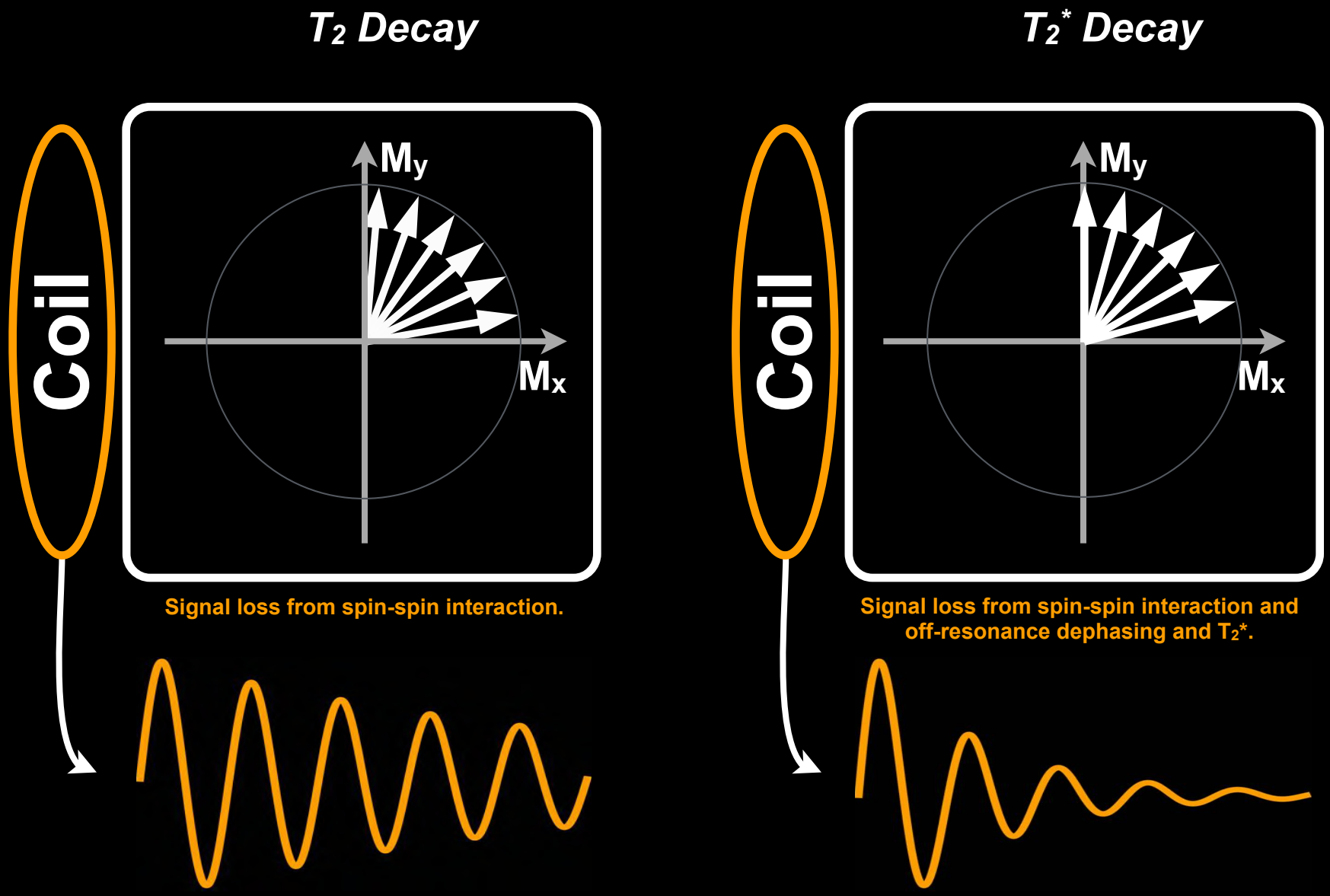
$$\frac{1}{T_2^*} = \frac{1}{T_2} + \frac{1}{T_2'} + \frac{1}{T_2^D} + \cdots$$

Irreversible Losses

Reversible Losses

Irreversible Losses

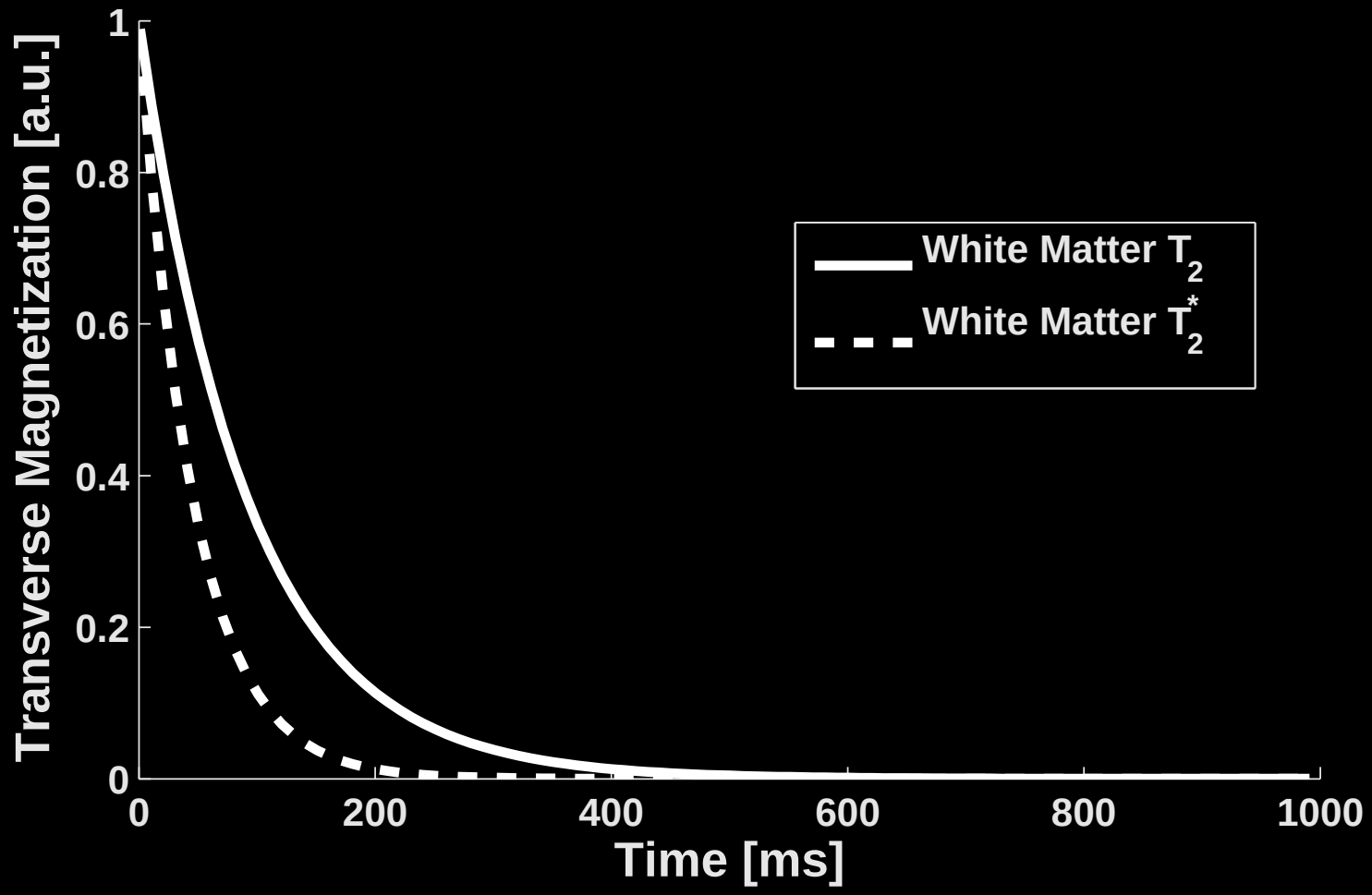
T_2 versus T_2^*



Signal loss from spin dephasing and T_2^*.

David Geffen School of Medicine
UCLA Radiology

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^*

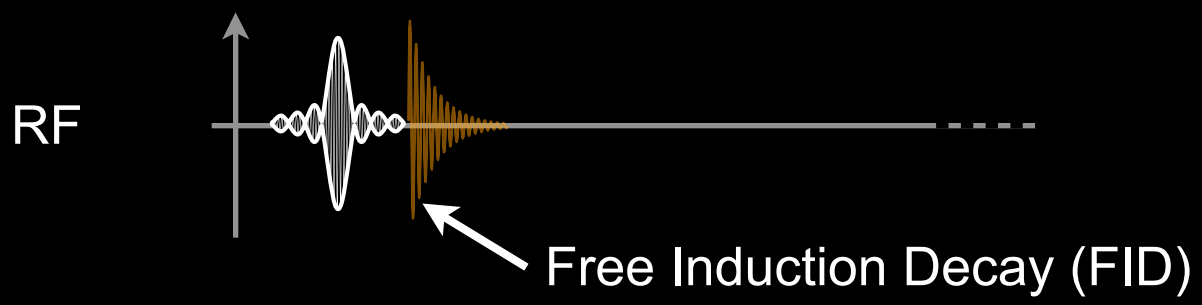
What are echoes?

What are echoes?

- **Two-sided NMR signals**
  - First half from re-focusing
  - Second half from de-phasing
- **Spin Echoes**
  - Arise from multiple RF-pulses
- **Gradient Echoes**
  - Arise from magnetic field gradient reversal
- **Line of *k*-space**

Why echoes?

- **Free Induction Decay**
  - NMR signal immediate after an RF pulse
  - Signal decays rapidly
    - T_2^*(<T_2)+Spectral distribution
- **Imaging requires certain "delays"**
  - Slice-selective re-phasing
  - Phase encoding
  - Readout pre-phasing
- **Echoes let us buy some time**



Pulse Sequence Definitions

- TR - Repetition Time
  - Duration of basic pulse sequence repeating block
  - At least one echo acquired per TR
- TE - Echo Time
  - Time from excitation to the maximum of the echo

Gradient Echo Imaging

Principal GRE Advantages

- **Fast Imaging Applications**
  - **Why?** *Can use a shorter TE/TR than spin echo.*
  - **When?** Breath-held, realtime, & 3D volume imaging

- **Bright blood signal**
  - **Why?** Inflowing spins haven't "seen" numerous RF pulses.
  - **When?** Cardiovascular & angiographic applications.

- **Low SAR**
  - **Why?** Imaging flip angles are small.
  - **When?** When heating risks are a concern (devices, high field)

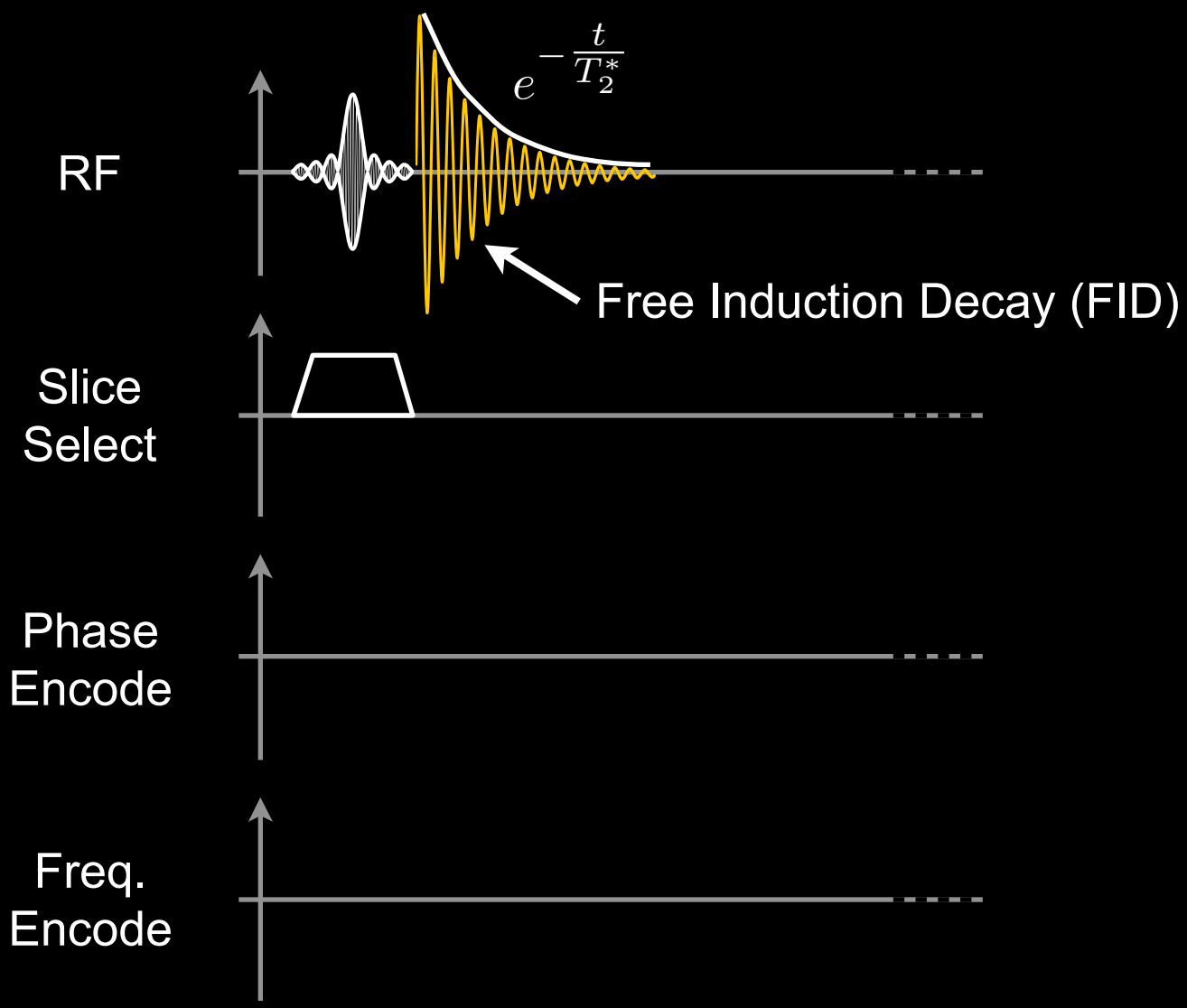
Principal GRE Advantages

- **Quantitative**
  - **Why?** Multi-echo acquisition are practical.
  - **When?** Flow quantification & Fat/Water mapping
- **Susceptibility Weighted Imaging**
  - **Why?** No refocusing pulse.
  - **When?** T_2^*-weighted & imaging hemorrhage
- **Reduced Cross-talk**
  - **Why?** SE hard to match slice profile of 90° & 180°
  - **When?** Little or no slice gap for 2D multi-slice

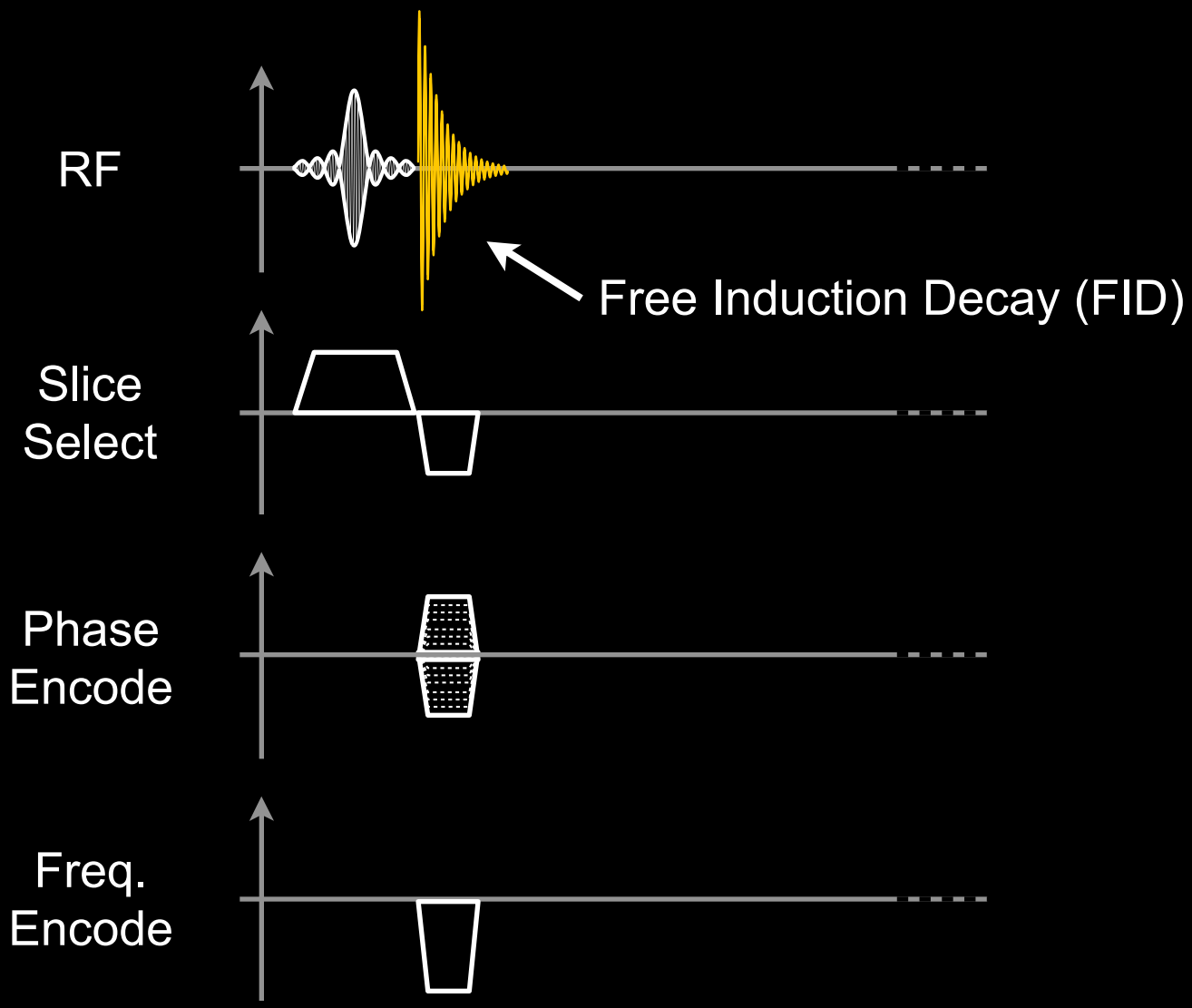
Principal GRE Disadvantages

- **Off-resonance sensitivity**
  - **Why?** Field inhomogeneity, Susceptibility, & Chemical shift
- **T_2^*-weighted rather than T_2-weighted**
  - **Why?** No re-focusing pulse
- **Larger metal artifacts than SE**
  - **Why?** No refocusing pulse.

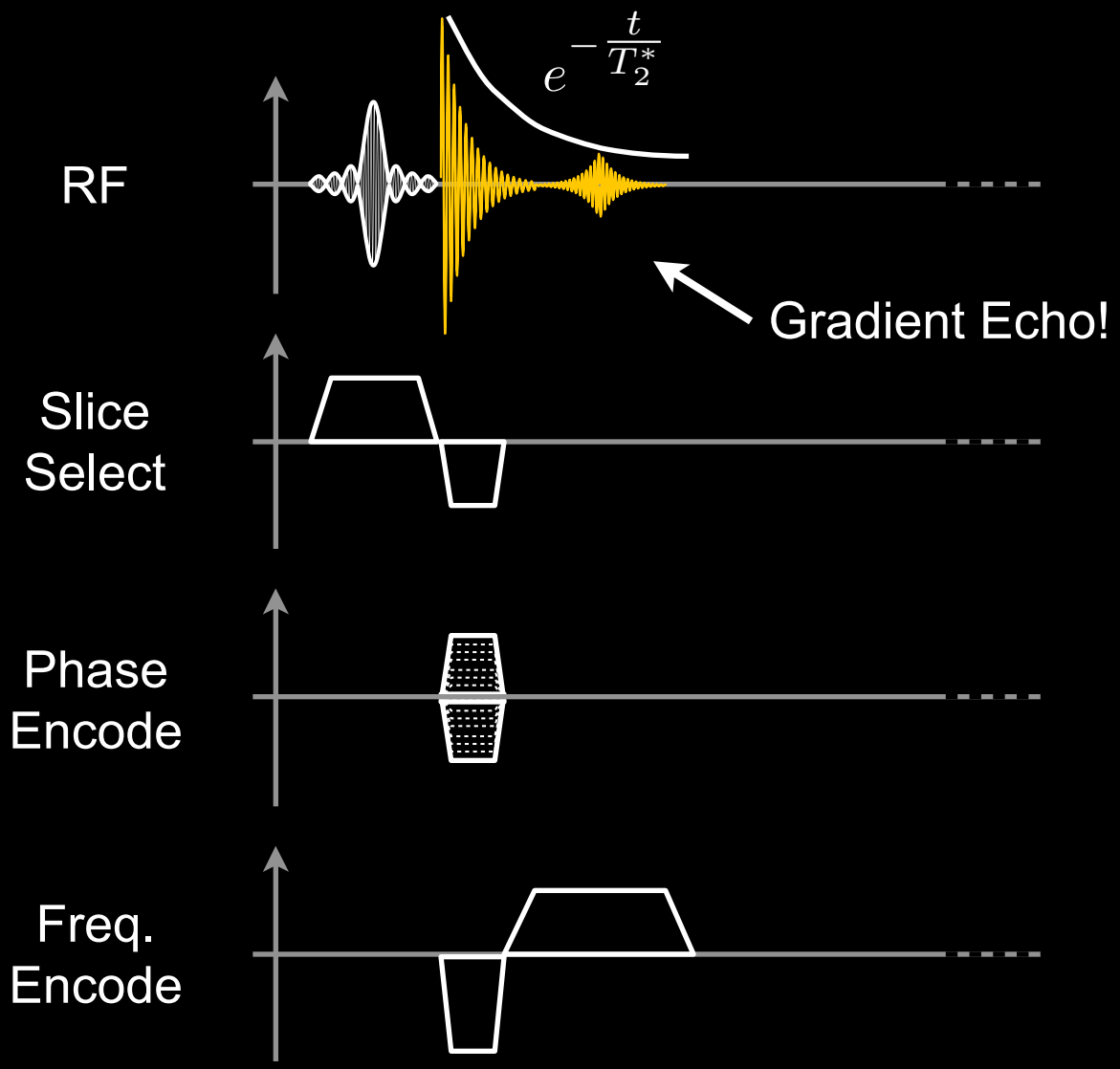
Basic Gradient Echo Sequence



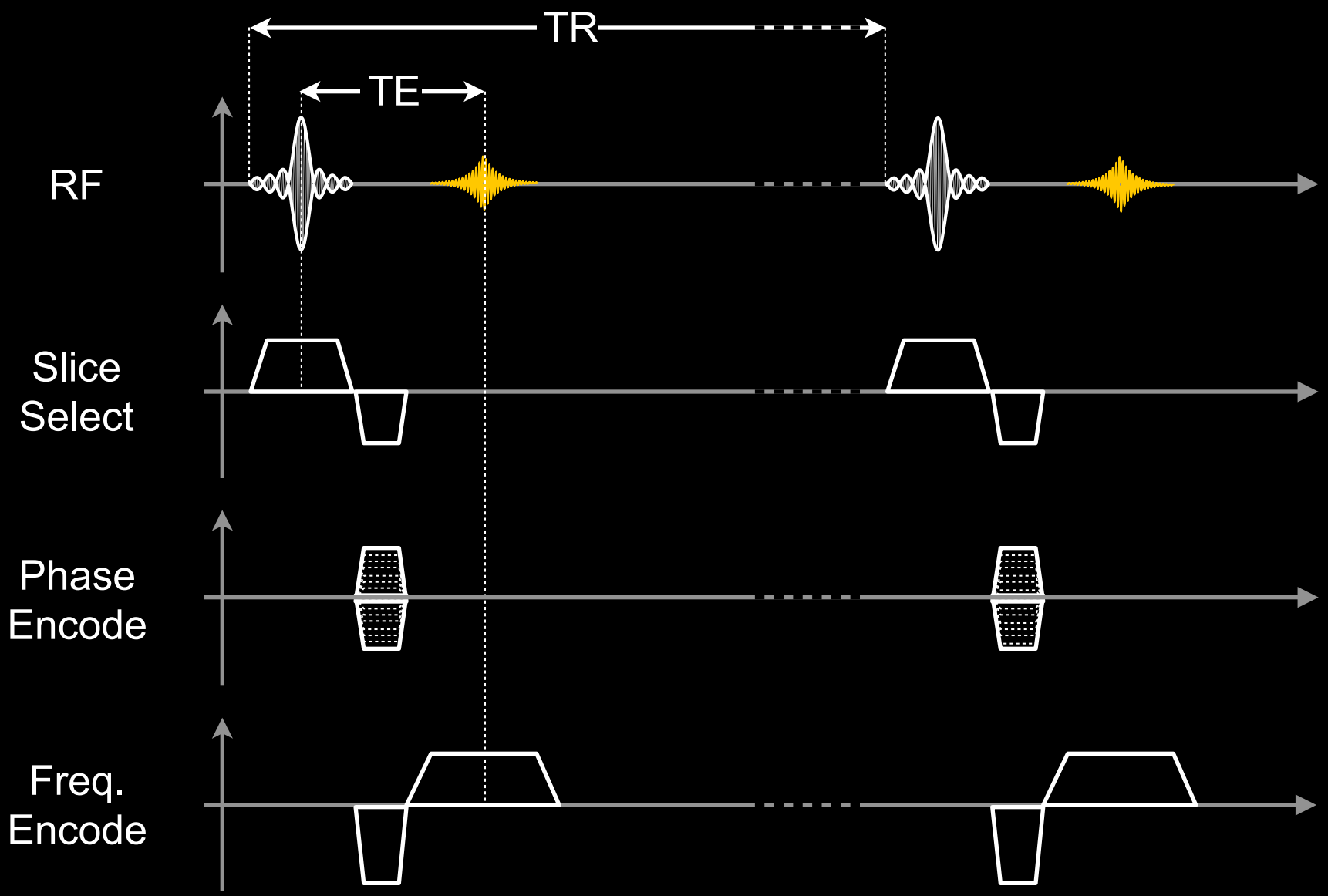
Basic Gradient Echo Sequence



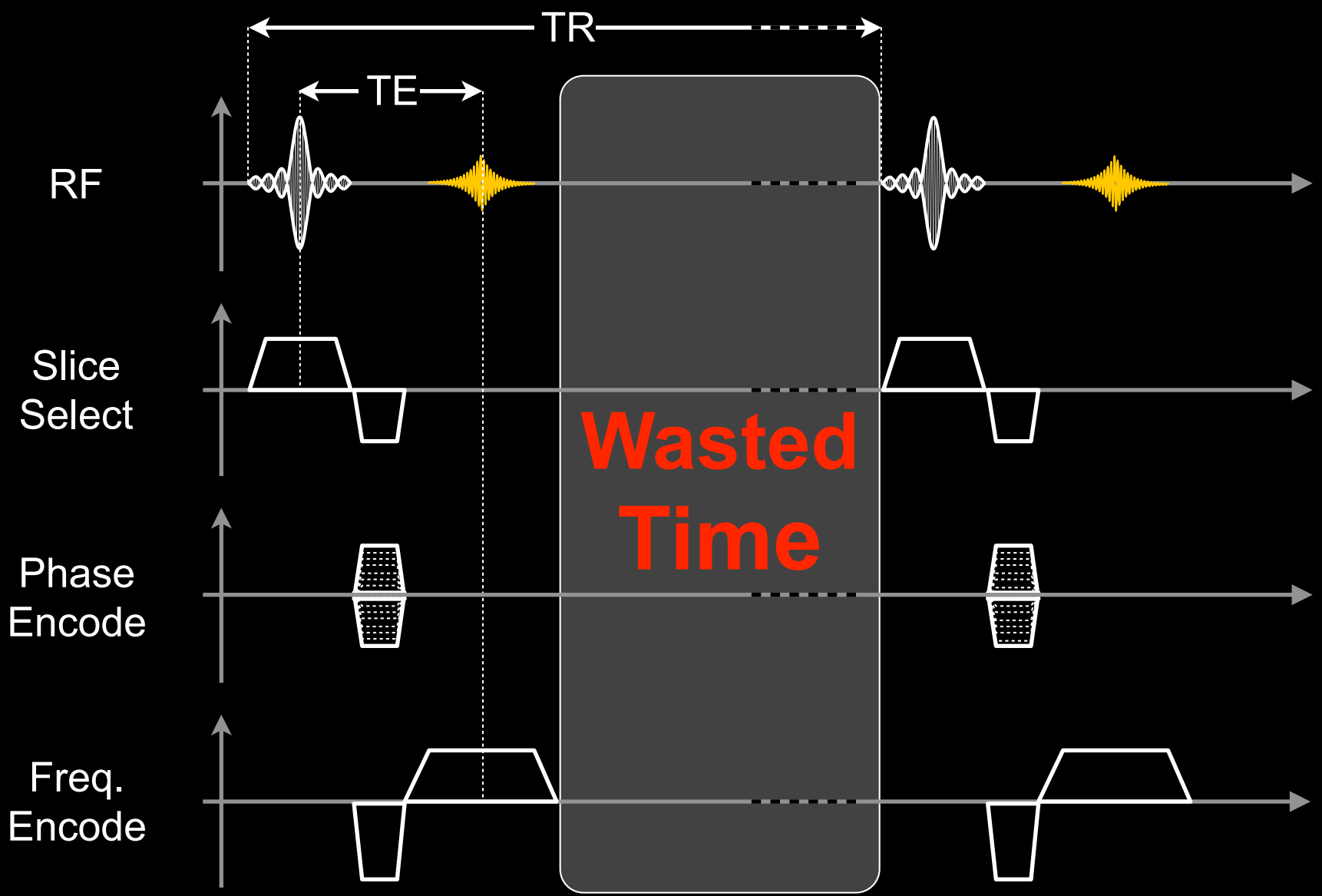
Basic Gradient Echo Sequence



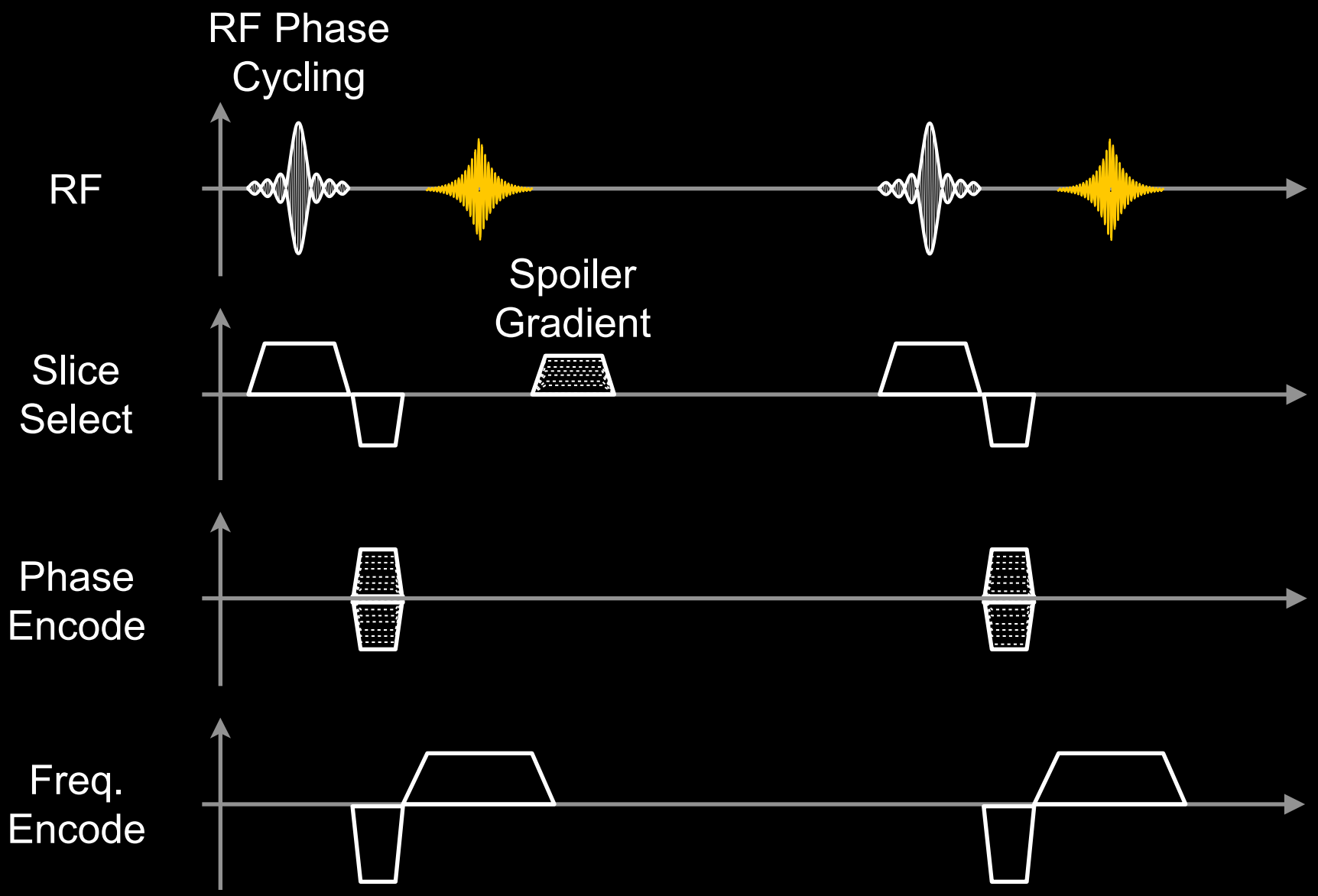
Basic Gradient Echo Sequence



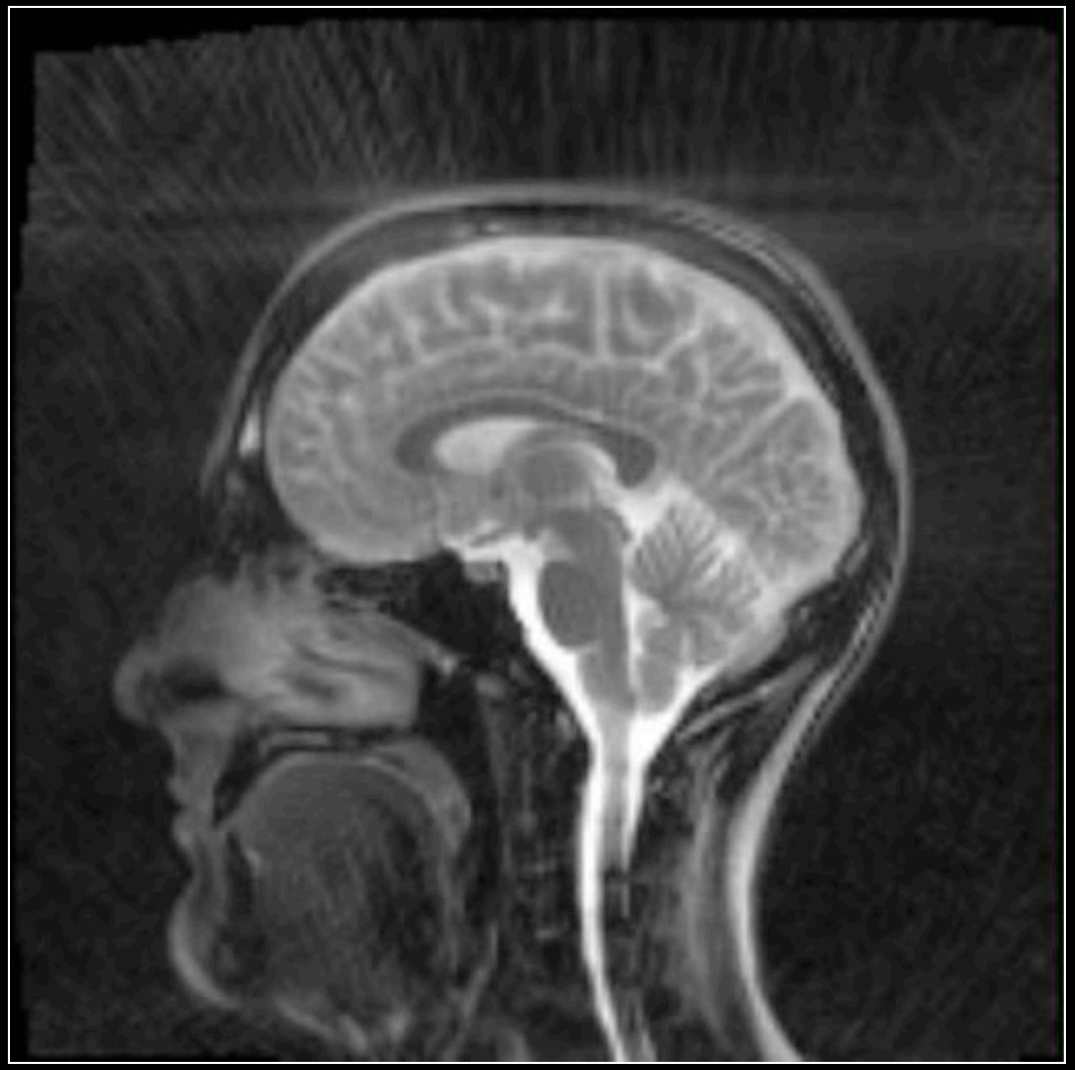
Basic Gradient Echo Sequence



Gradient Echo + Spoiling



Realtime Imaging with Gradient Echoes



Realtime imaging requires very short TE/TRs for rapid image acquisition.

Gradient Echoes & Contrast

Spoiled Gradient Echo Contrast

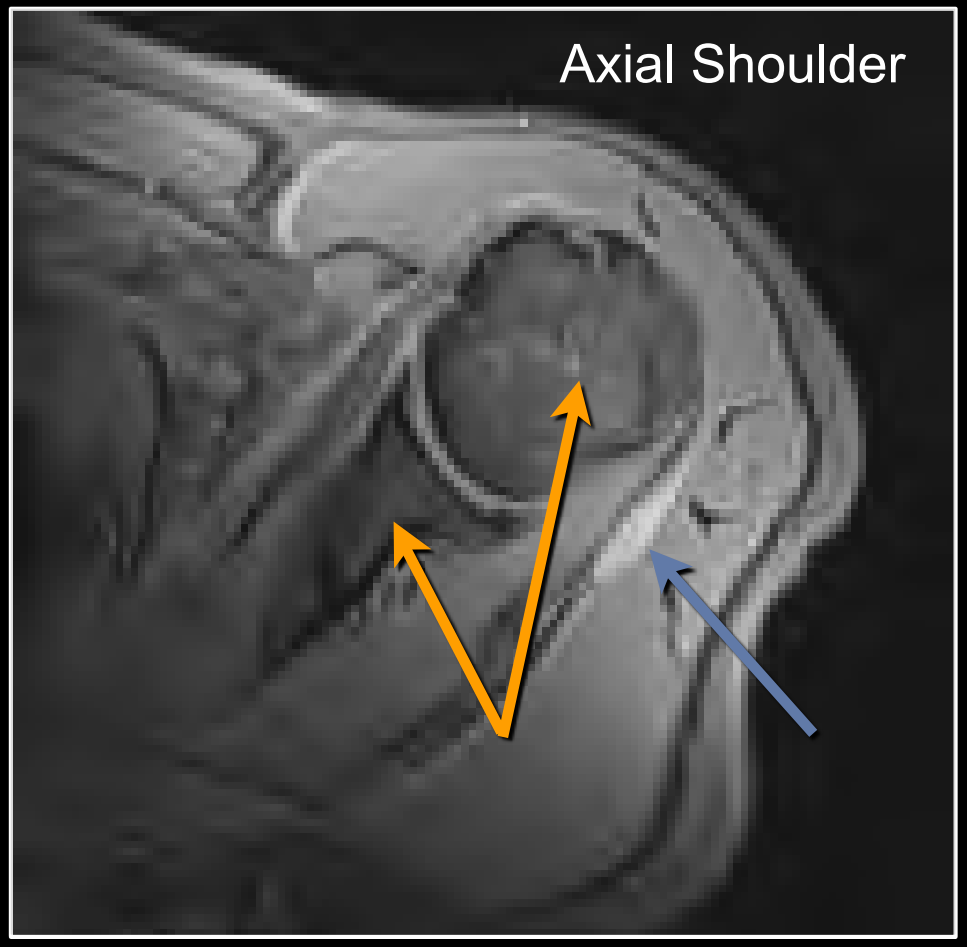
Gradient Echo Parameters

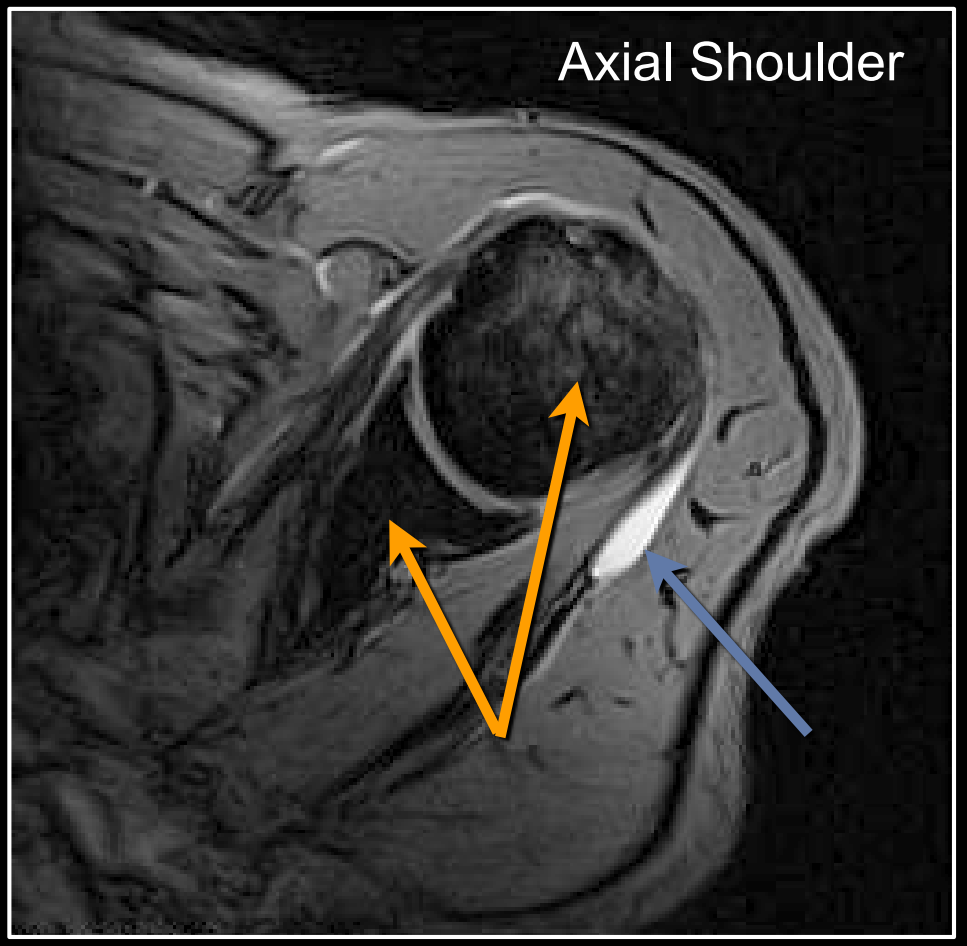
Type of Contrast	TE	TR	Flip Angle
Spin Density	Short	Long	Small
T_1-Weighted	Short	Intermediate	Large
T_2^*-Weighted	Intermediate	Long	Small

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

Contrast adjusted by changing TR, flip angle, and TE.

T_2*-weighted Gradient Echo Imaging





TE=9ms **TE=30ms**

Susceptibility Weighting (darker with longer TE)

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

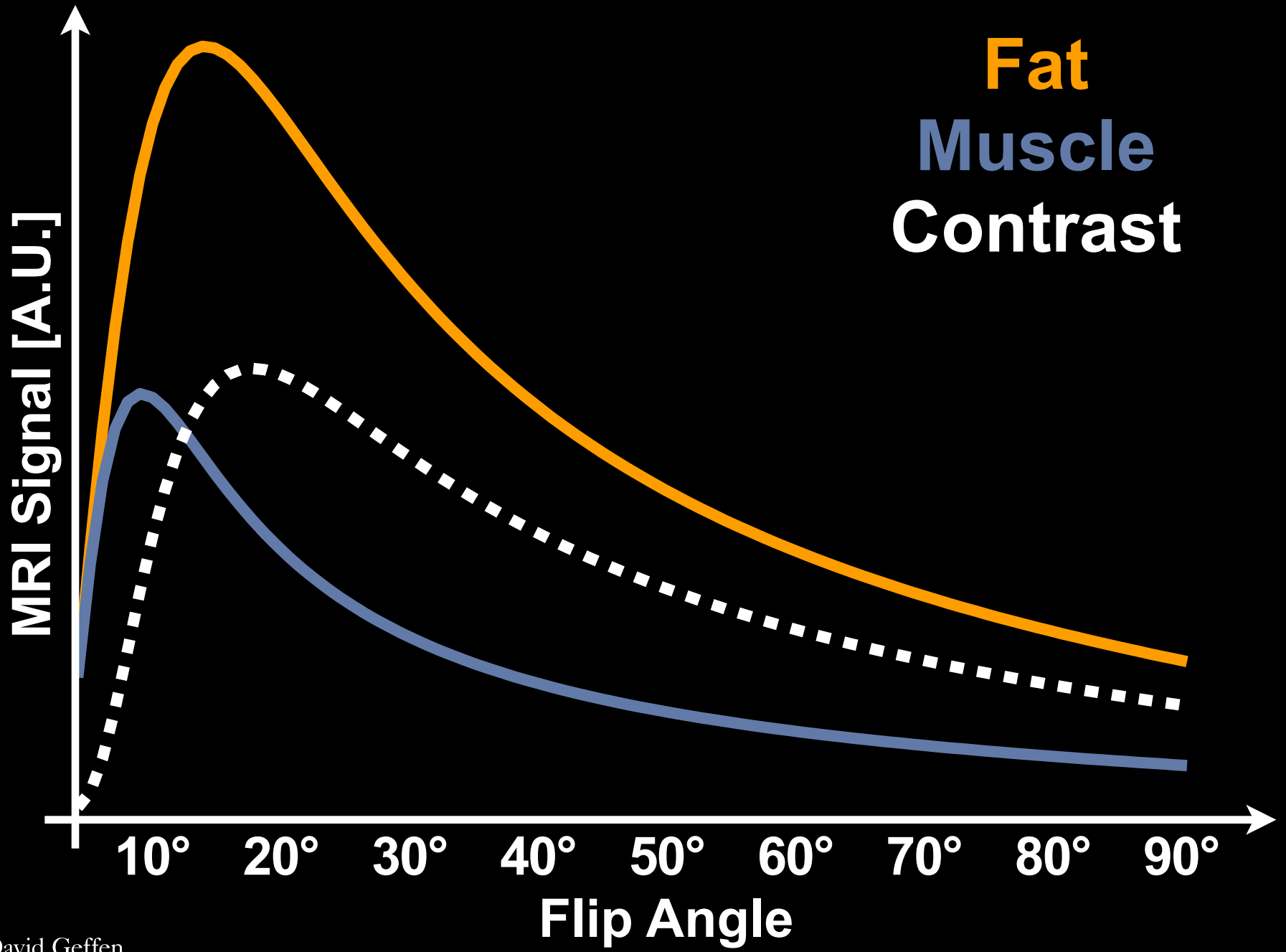
Spoiled GRE & Ernst Angle

$$\alpha_{Ernst} = \arccos\left(e^{-\frac{TR}{T_1}}\right)$$

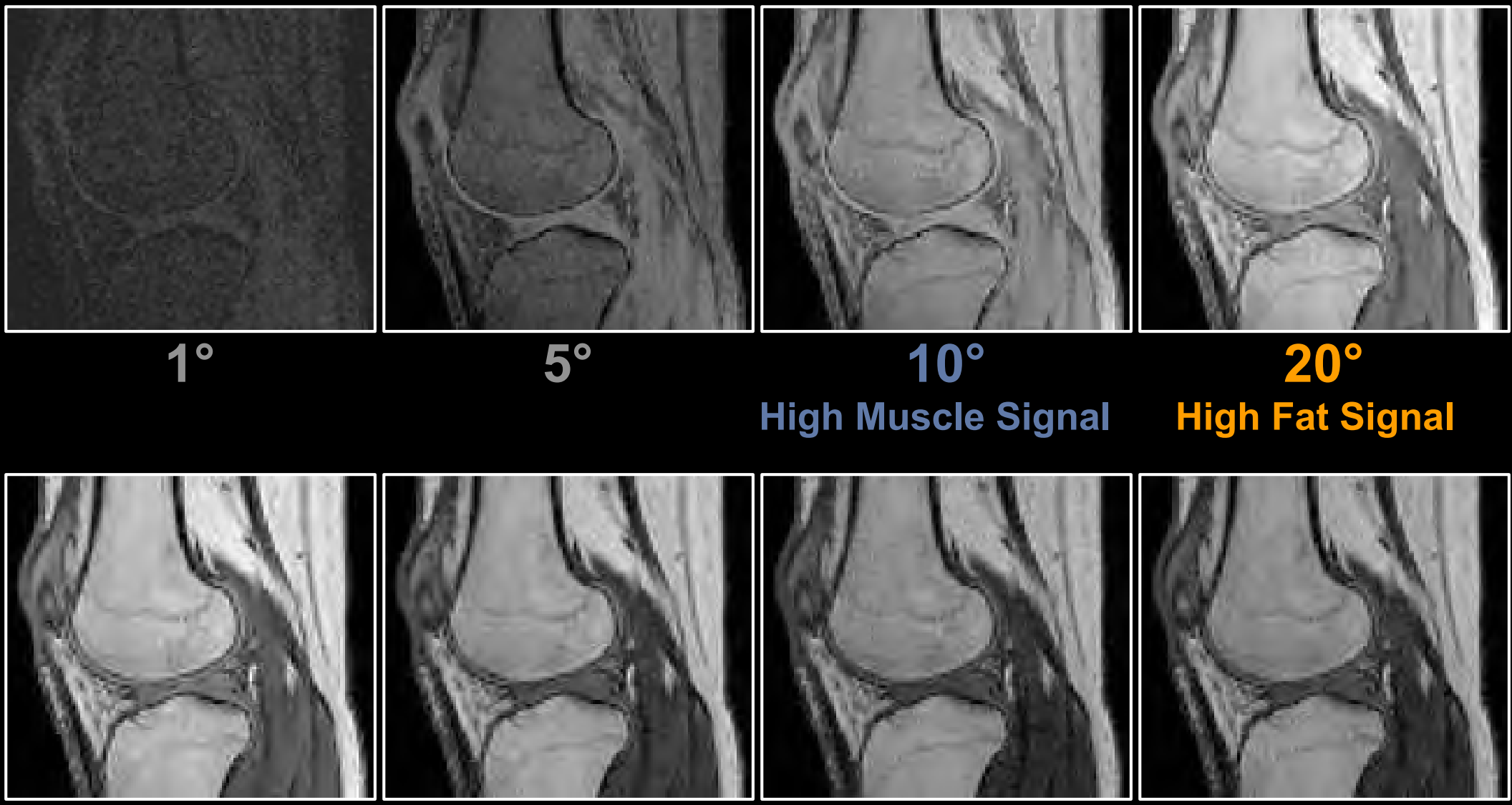
Produces the largest MRI signal for a given TR and T_1.

Tissue	T_1 [ms]	T_2 [ms]
muscle	875	47
fat	260	85

Spoiled GRE & Ernst Angle



Spoiled GRE & Ernst Angle



1°

5°

10°
High Muscle Signal

20°
High Fat Signal

30°
Highest Contrast

45°

60°

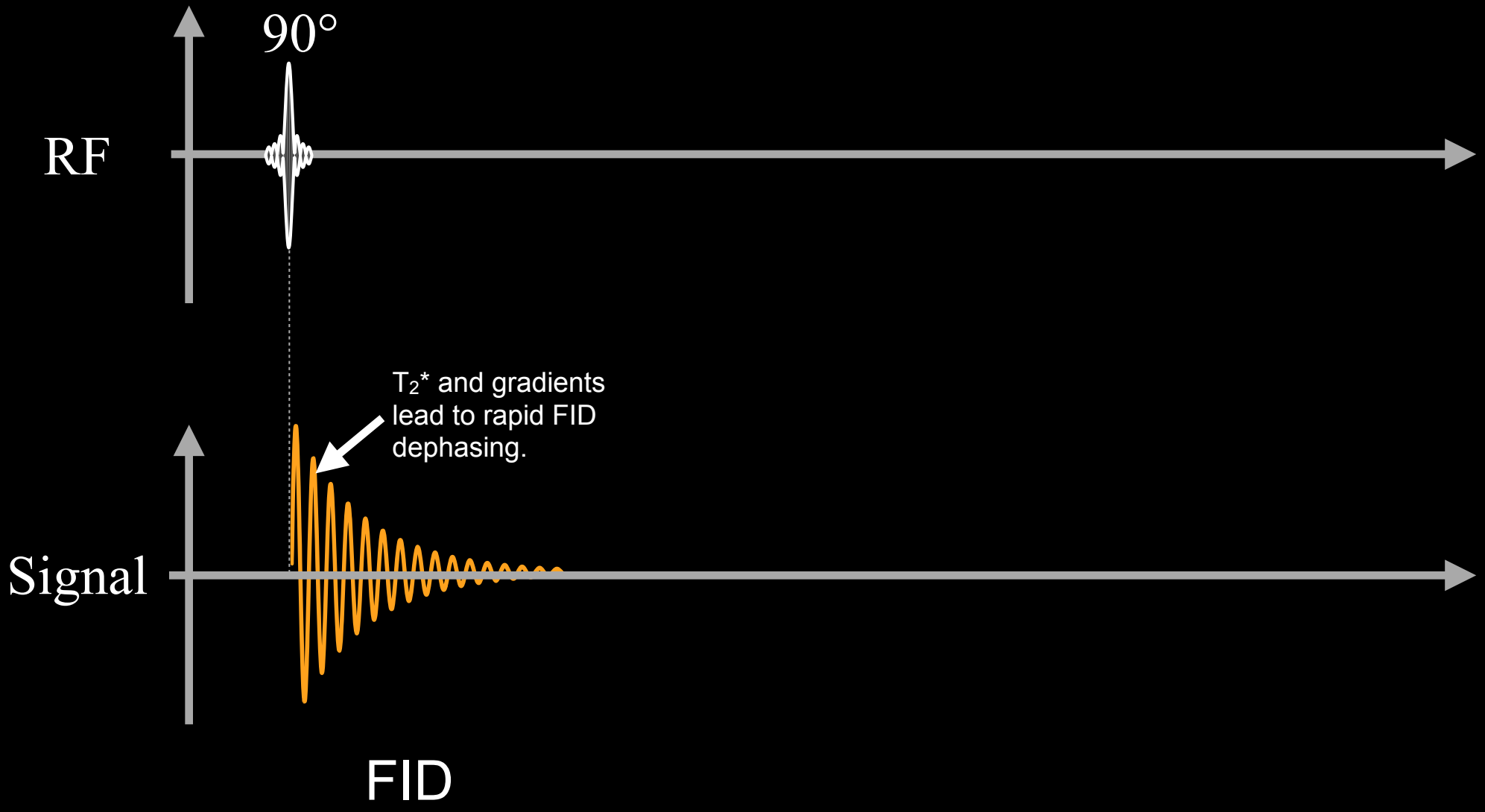
90°

Spin Echo Imaging

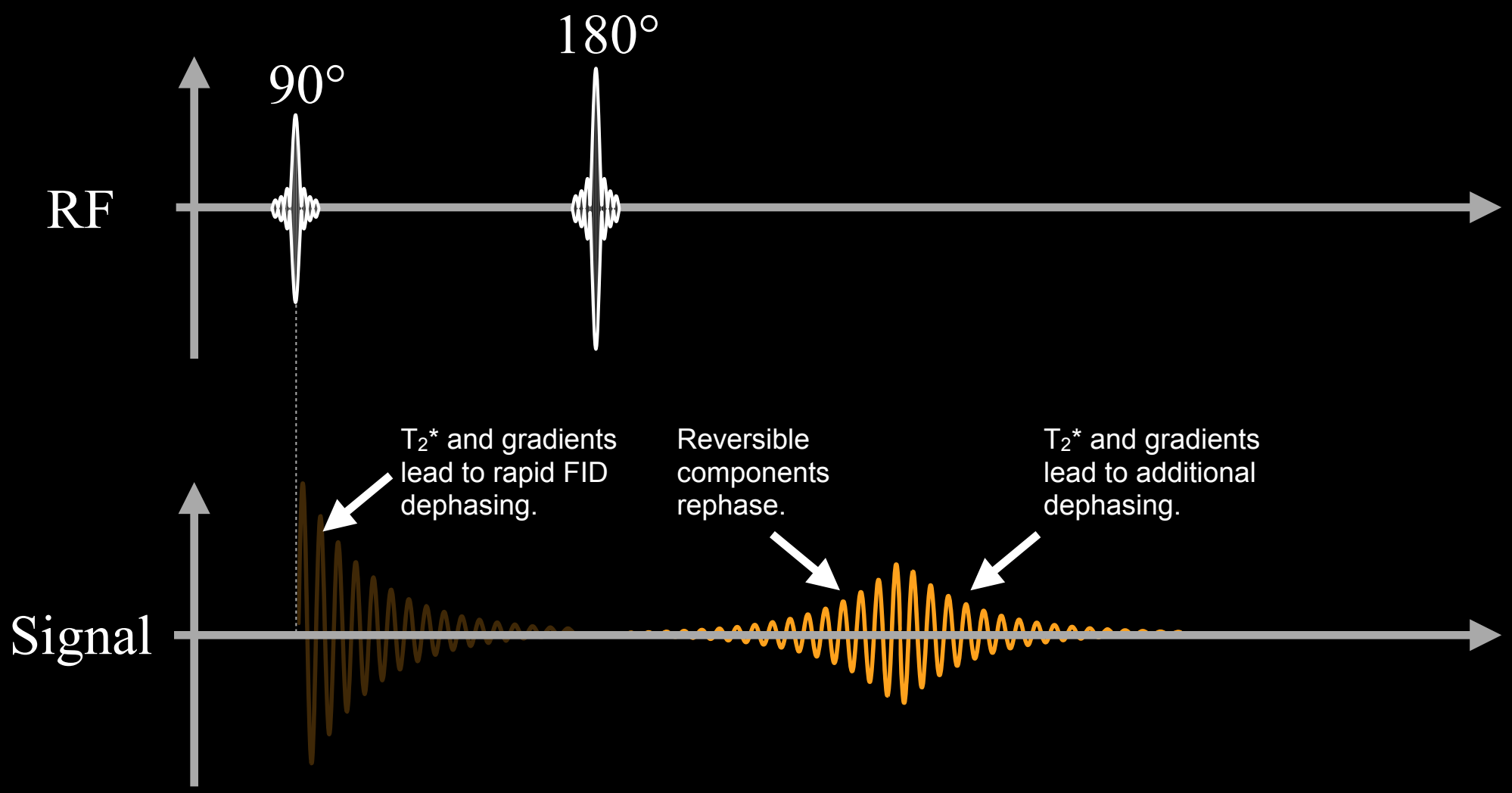
Spin Echo

- **Advantages**
  - **All spins within voxel rephased**
    - **Insensitive to off-resonance**
      - **B_0 inhomogeneity**
      - **Intravoxel Chemical shift signal loss**
      - **Susceptibility**
    - **Off-resonance causes spin-dephasing**
  - **Great for T_1, T_2, ρ contrast**
    - **Not T2***
  - **High SNR**
- **Disadvantages**
  - **TR can be long**
  - **SAR can be high**

Free Induction Decay



Spin Echo

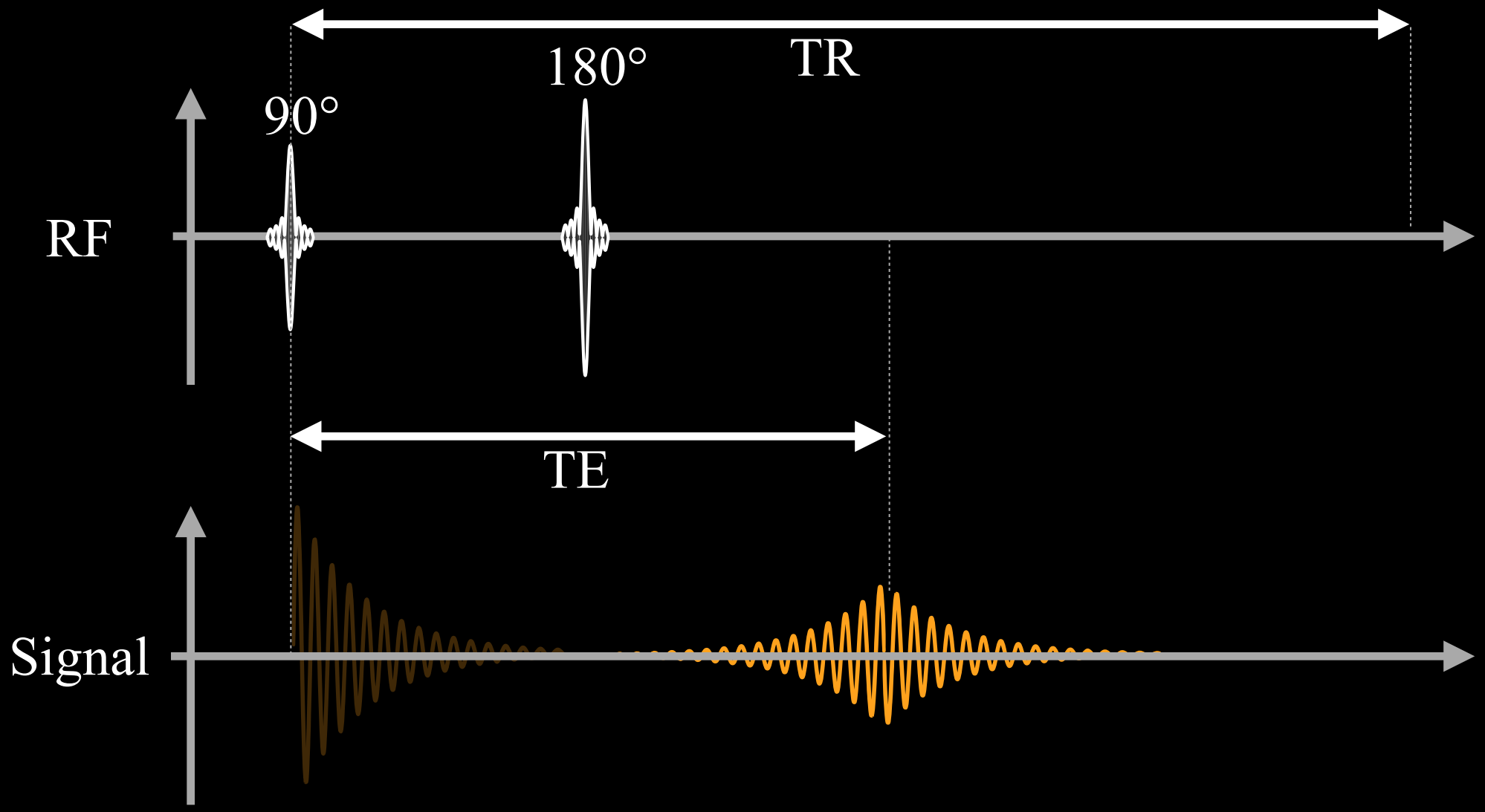


FID

Spin Echo

Record by computer.

Spin Echo

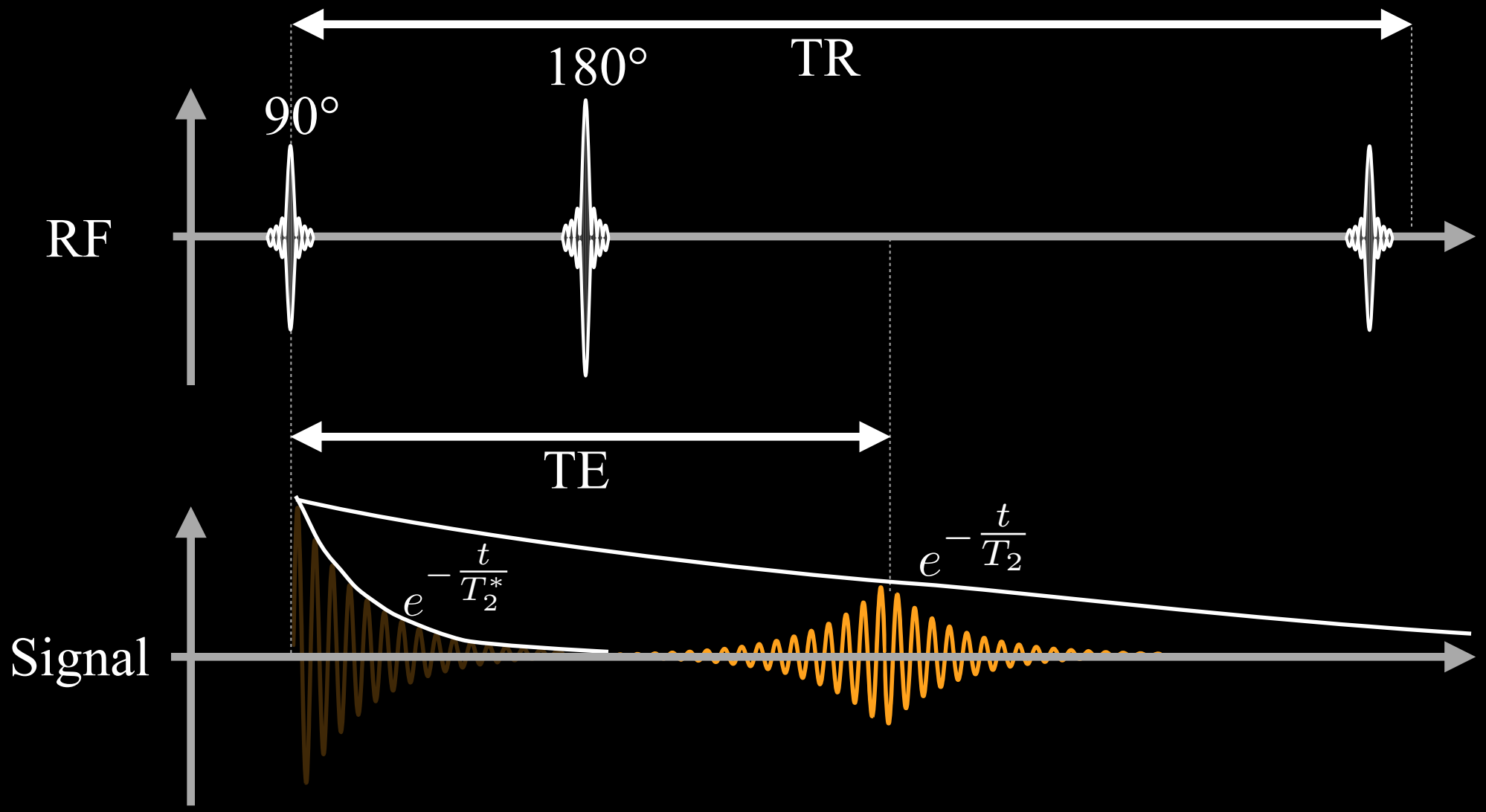


FID

Spin Echo

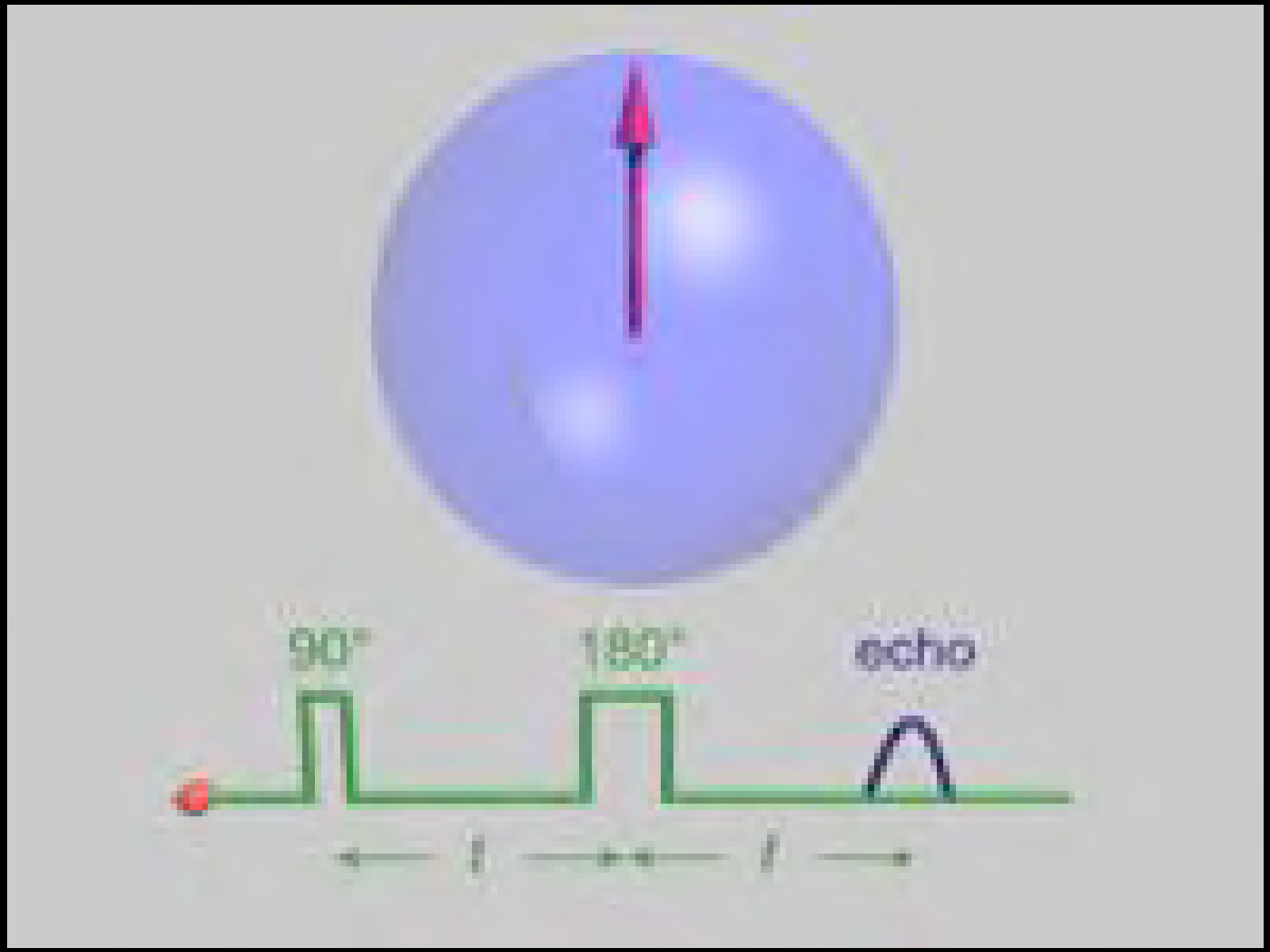
Record by computer.

Spin Echo - Contrast



How do you adjust the TR?
How do you adjust the TE?

Spin Echo - Refocusing



http://en.wikipedia.org/wiki/File:HahnEcho_GWM.gif

Spin Echoes & Contrast

Spin Echo Contrast

Spin Echo Parameters

	TE	TR
Spin Density	Short	Long
T_1-Weighted	Short	Intermediate
T_2-Weighted	Intermediate	Long

$$A_{Echo} \propto \rho \left(1 - e^{-TR/T_1}\right) e^{-TE/T_2}$$

Longer TR Minimizes This Term

Short TE Maximizes This Term

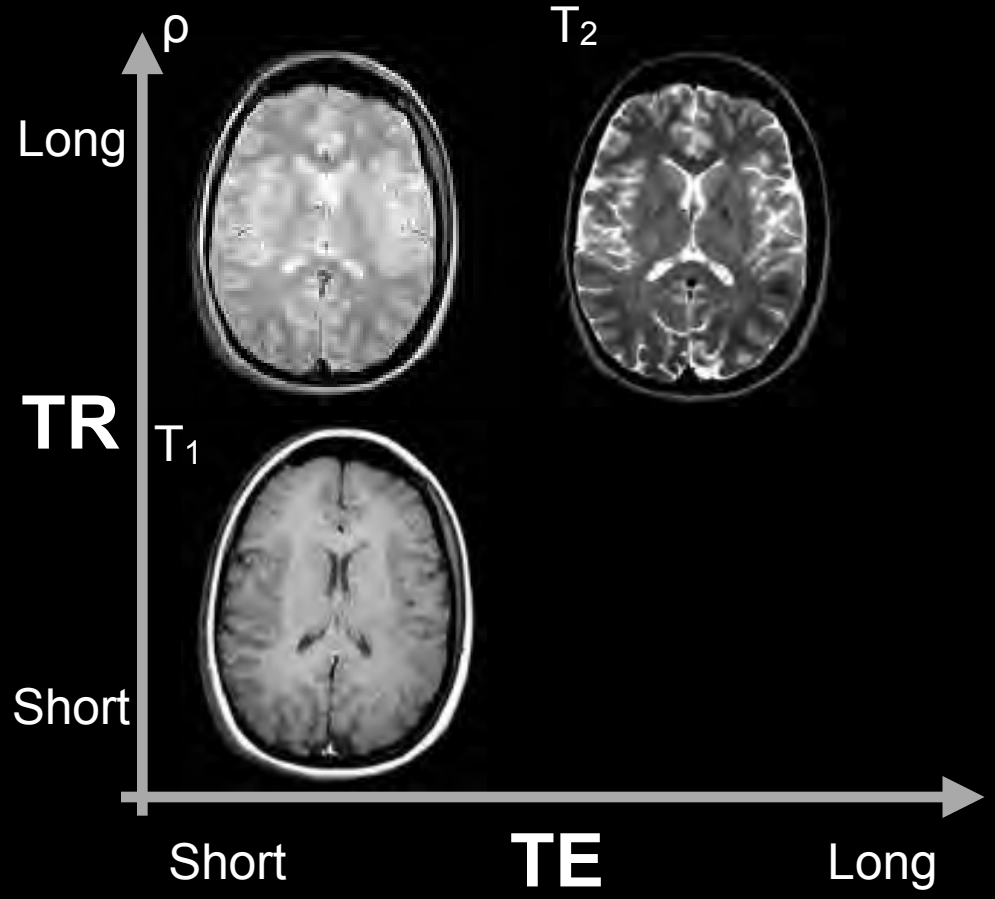
Short T_1 Maximizes This Term

Long T_2 Maximizes This Term

Spin Echo Contrast

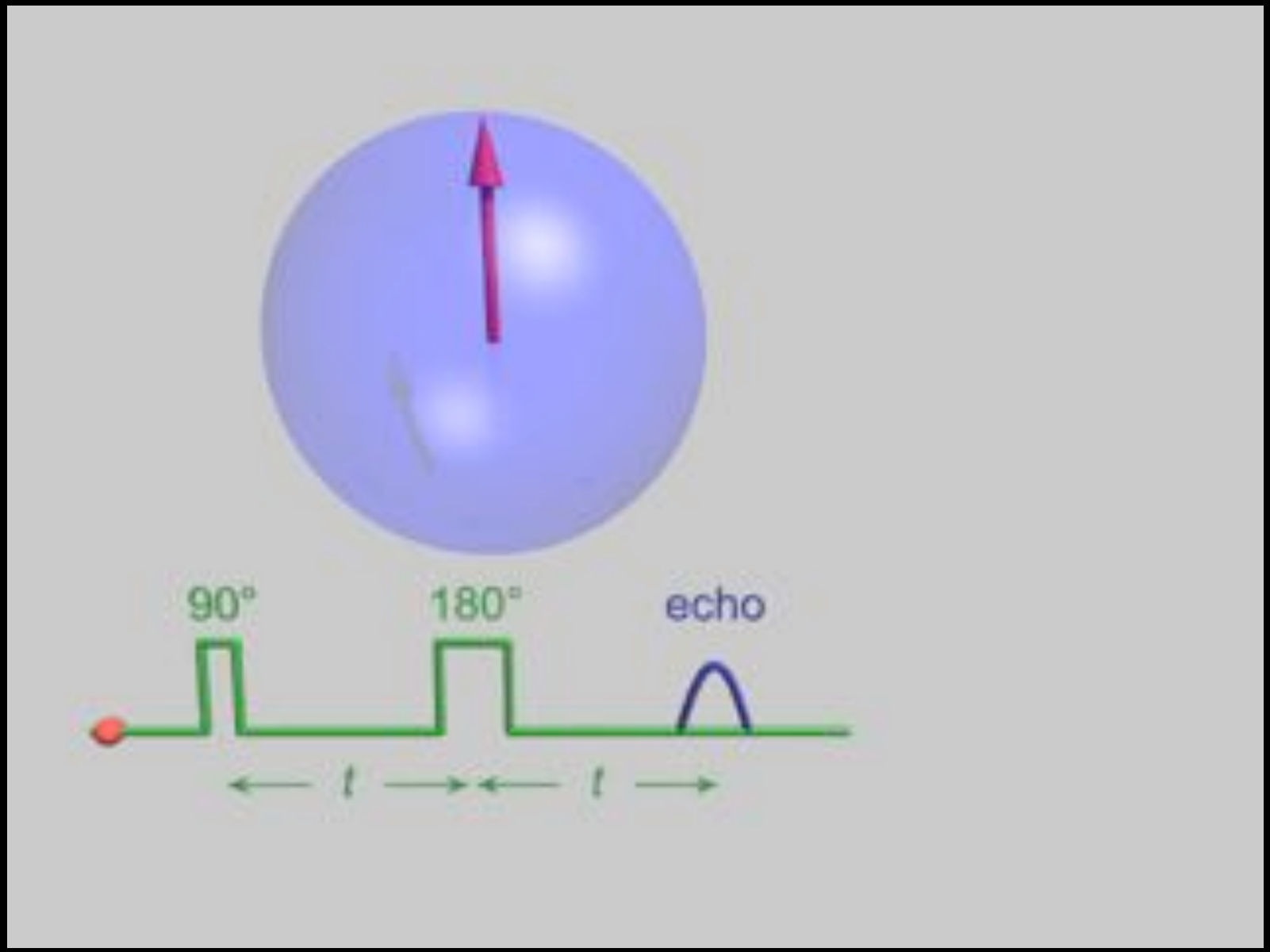
Spin Echo Parameters

	TE	TR
Spin Density	Short	Long
T_1-Weighted	Short	Intermediate
T_2-Weighted	Intermediate	Long



Images Courtesy of Mark Cohen

Spin Echo - Contrast

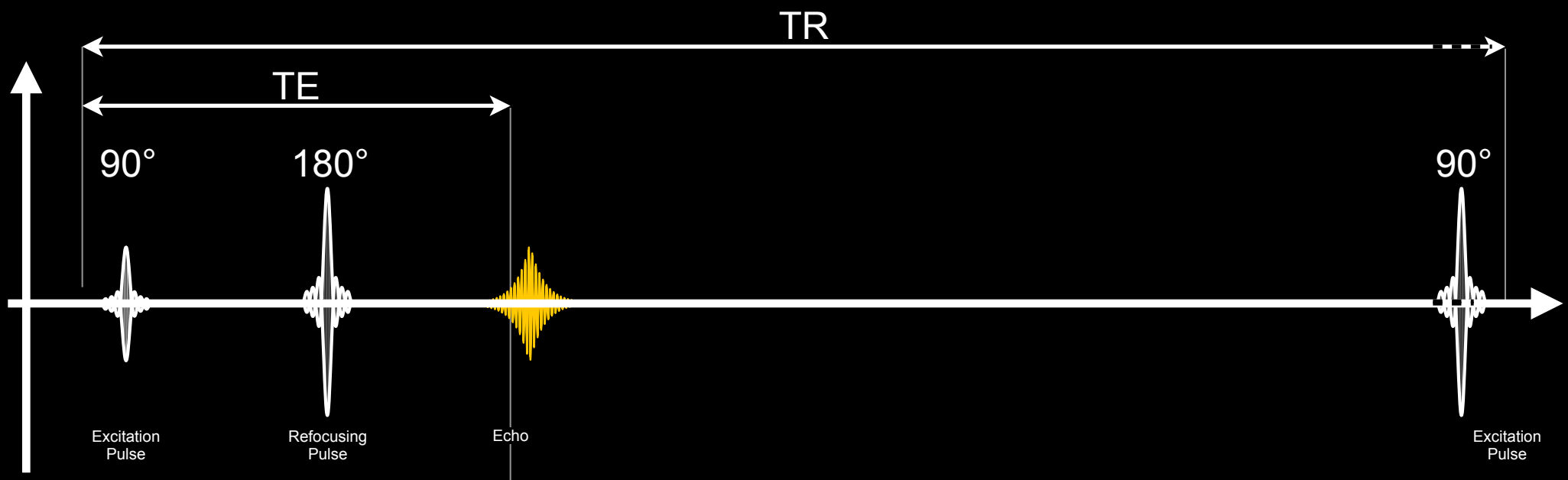


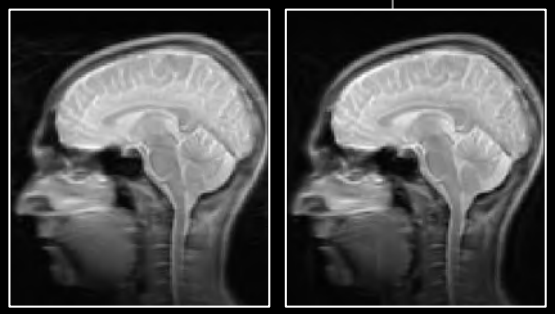
http://en.wikipedia.org/wiki/File:HahnEcho_GWM.gif

Spin Echo

TR
TE
90°
180°
90°
Excitation Pulse
Refocu Pulse
Echo
Excitation Pulse
TE=12ms

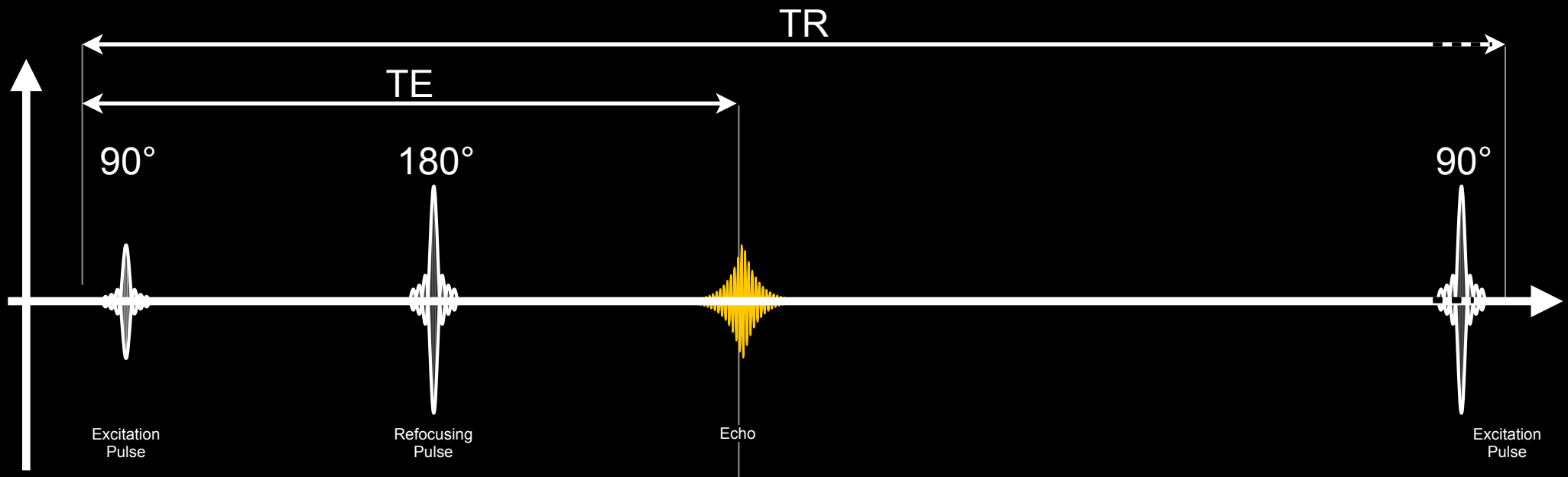
Spin Echo

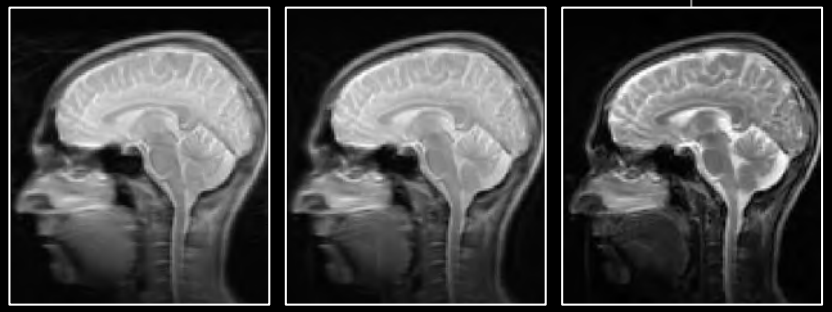




TE=12ms TE=47ms

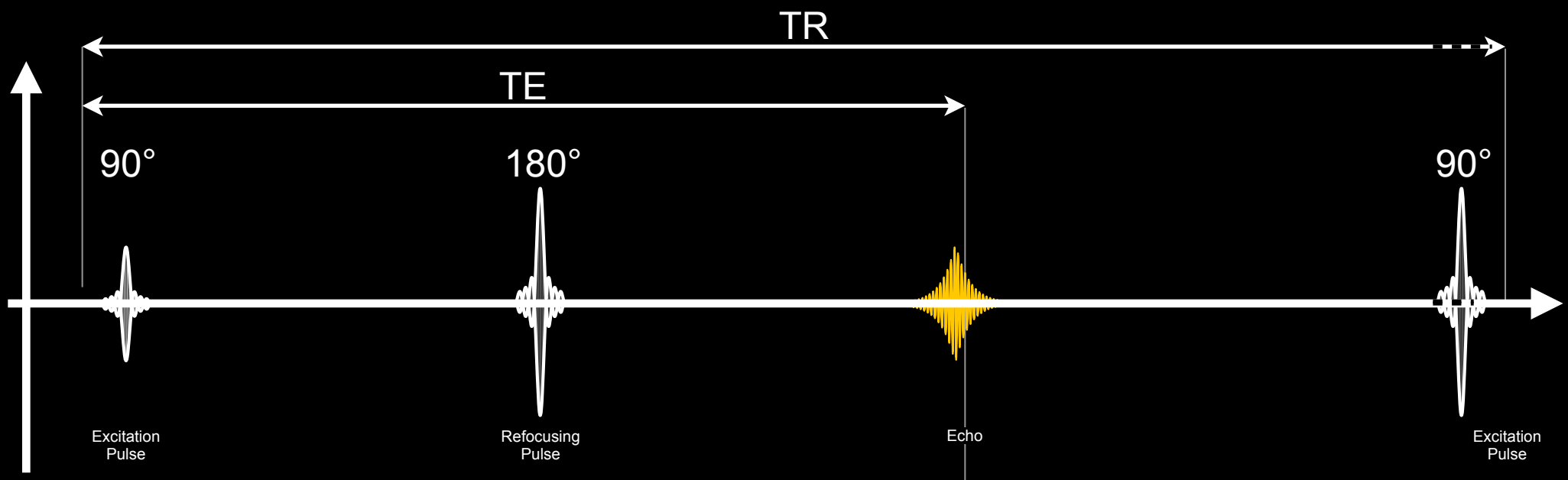
Spin Echo

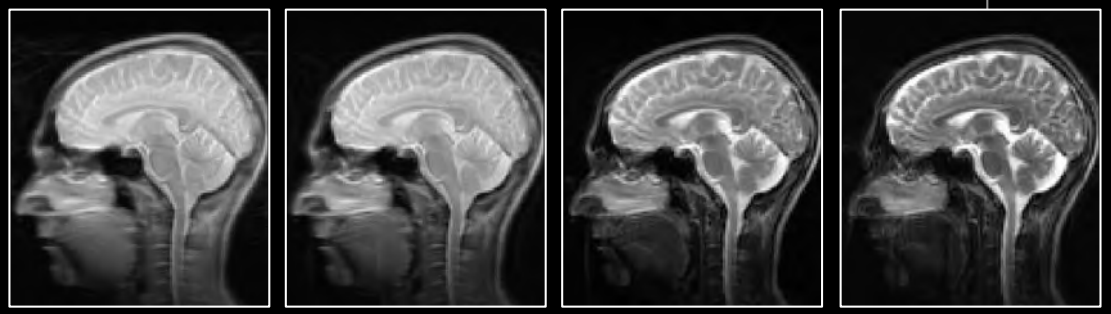




TE=12ms TE=47ms TE=106ms

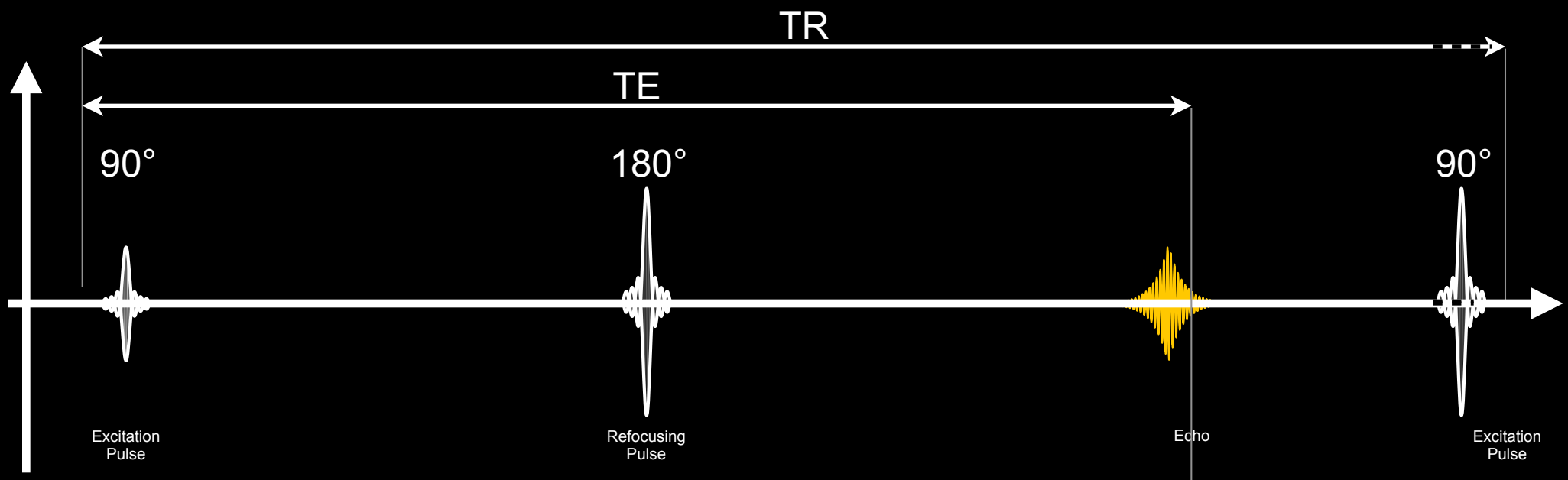
Spin Echo

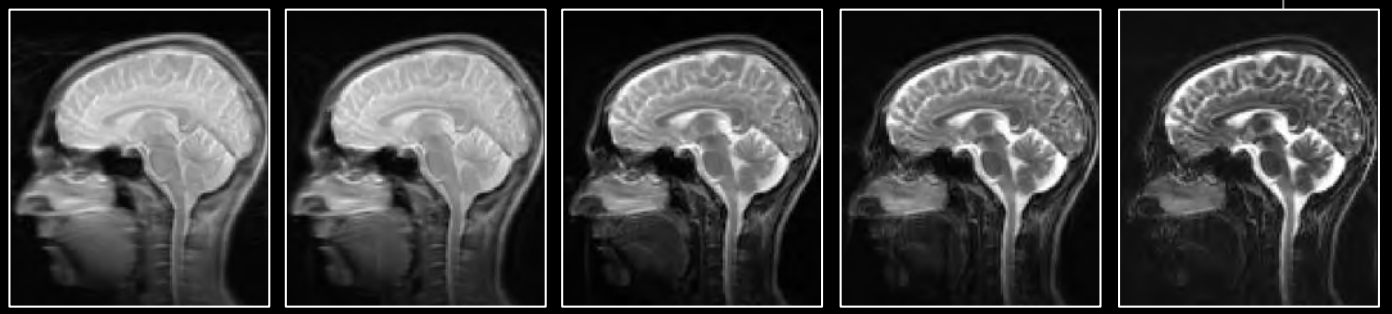




TE=12ms TE=47ms TE=106ms TE=153ms

Spin Echo





TE=12ms TE=47ms TE=106ms TE=153ms TE=235ms

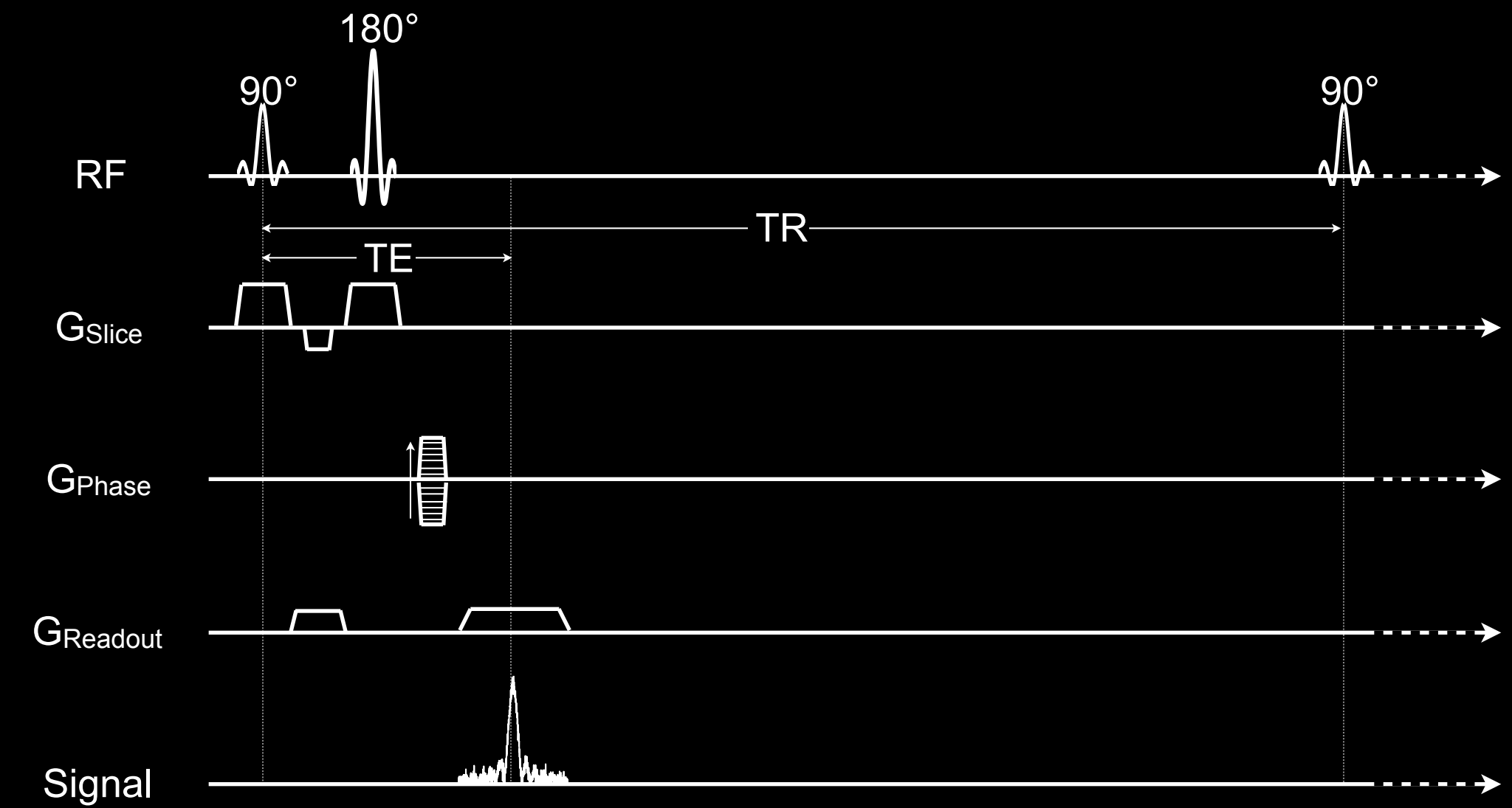
Multi-Echo Imaging

How do we calculate scan time?

$$T_{Scan} = TR \cdot PE \cdot N_{avg}$$

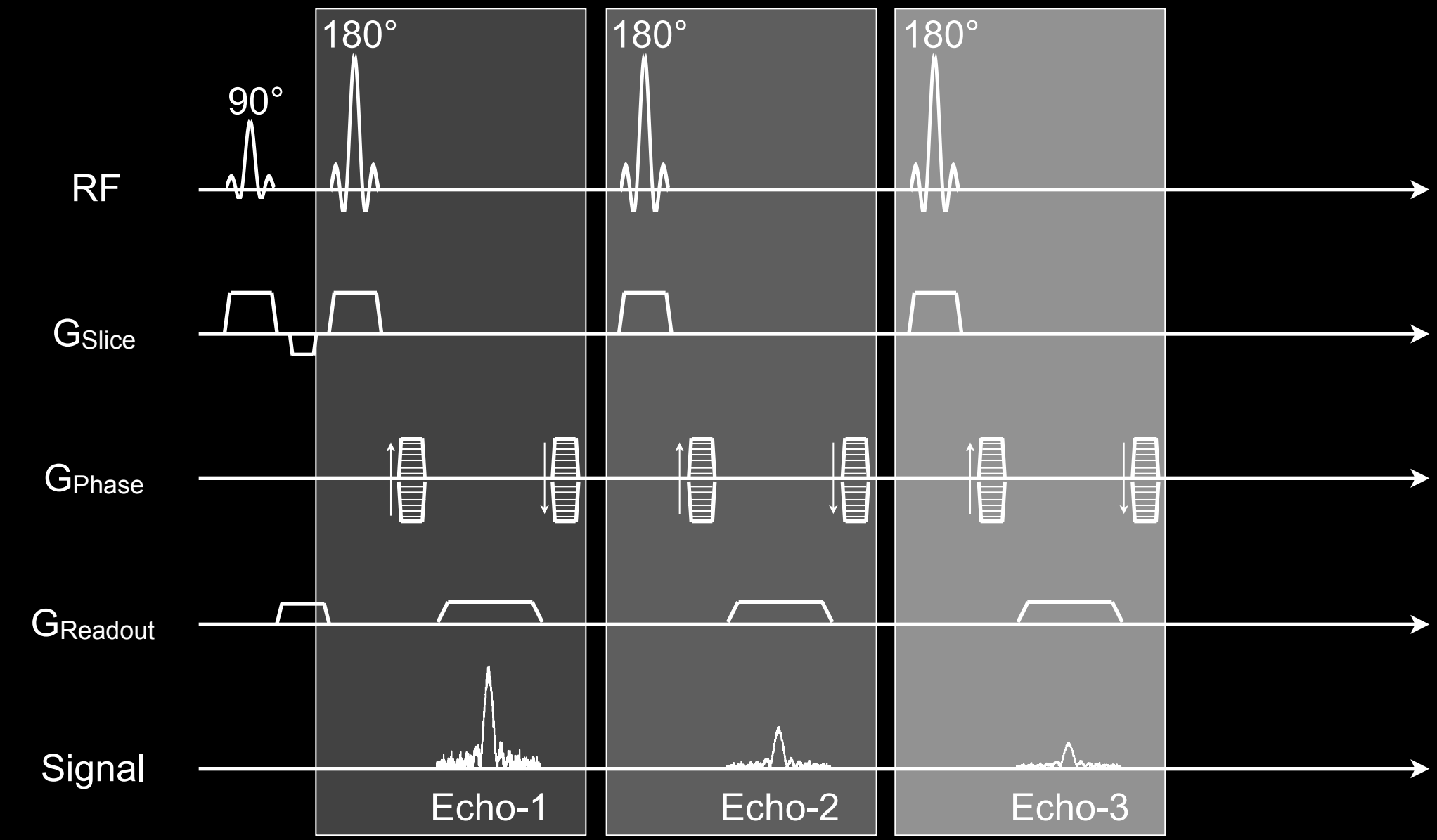
- T_{Scan}=1000ms•256•1=4:16 [mm:ss]
- Assumes one echo per TR.
- PE - Number of phase encode lines (N_{ky})
- N_{Avg} - Number of signal averages

Spin Echo

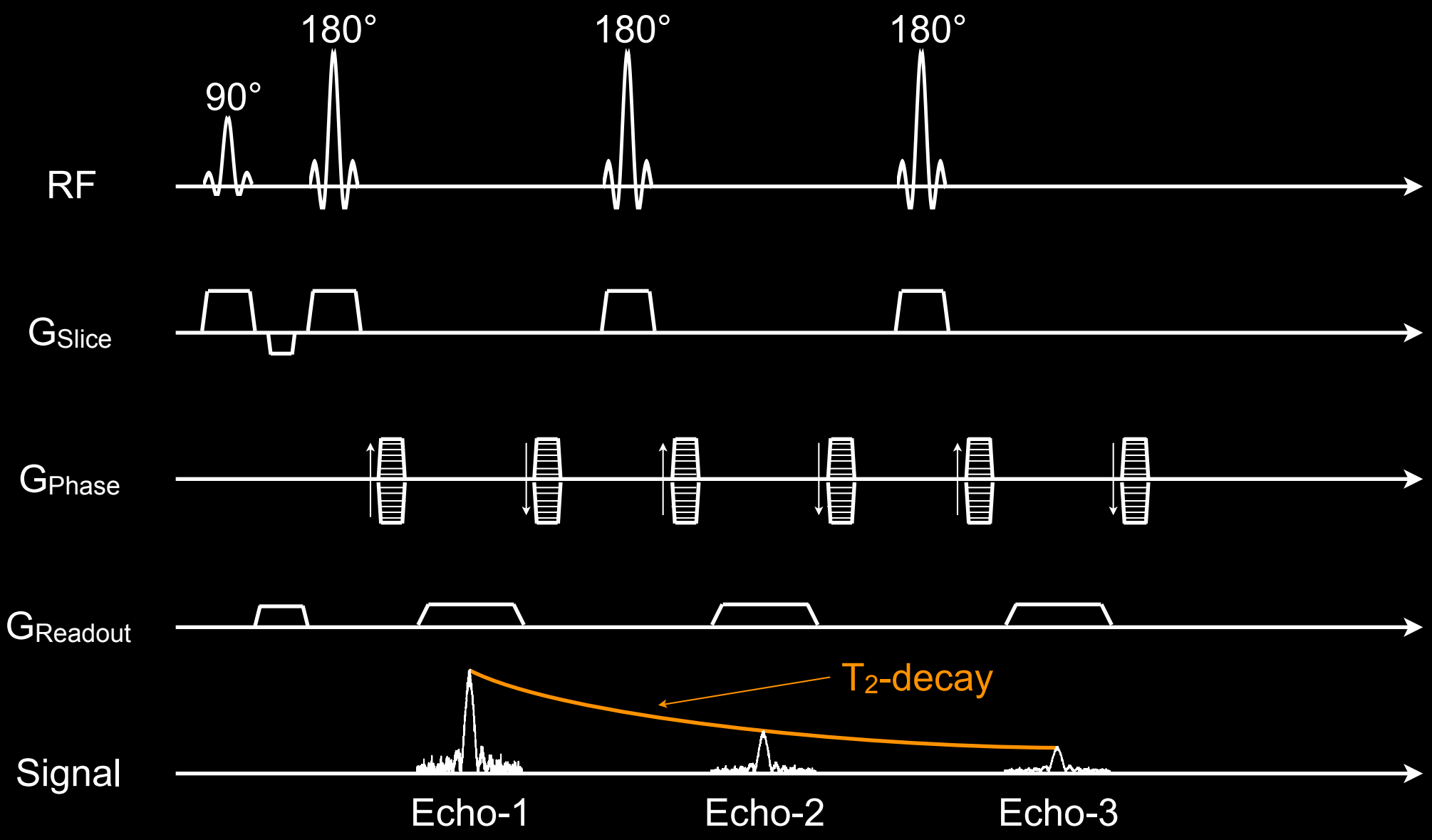


Spin Echo

Fast Spin Echo



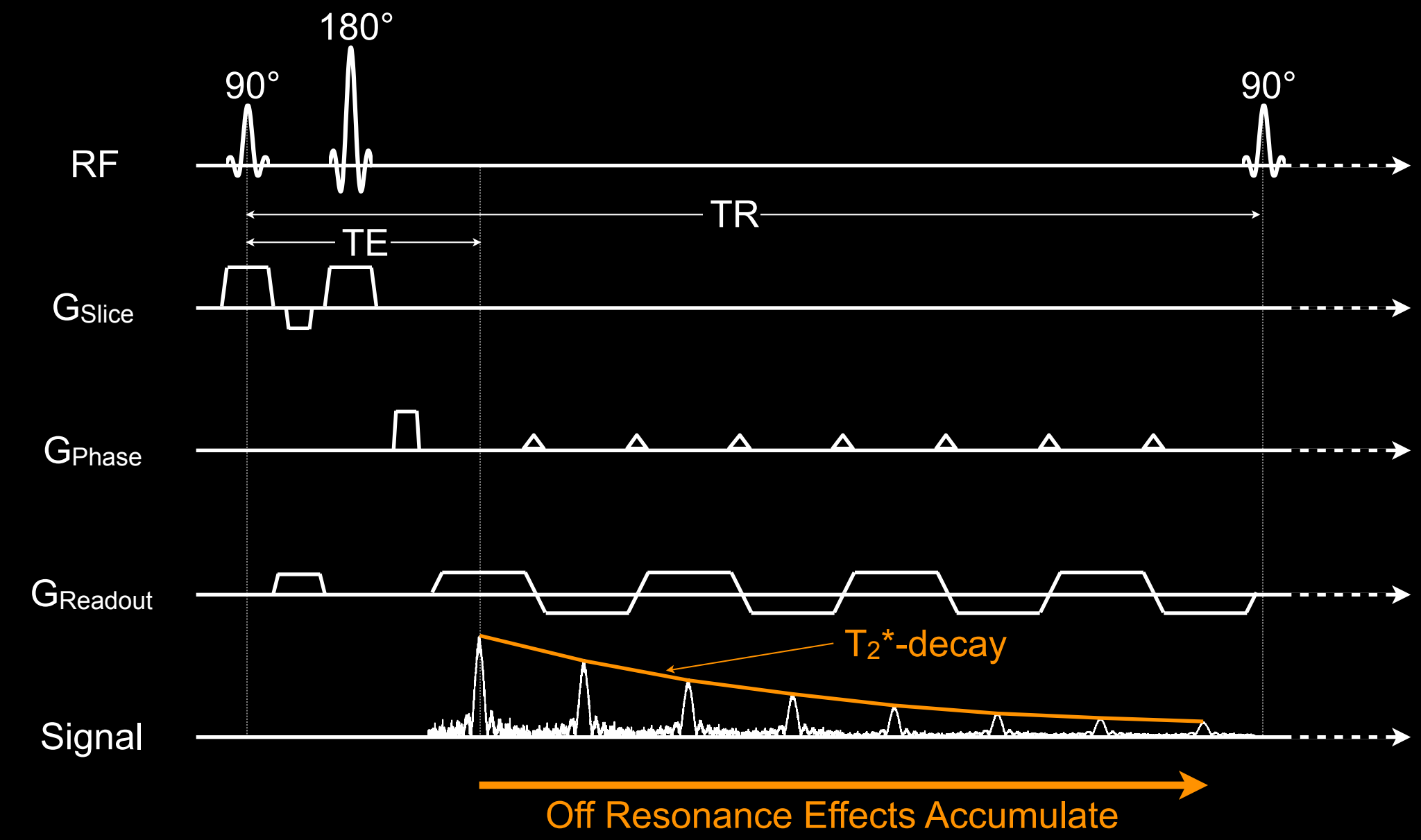
Fast Spin Echo



Fast Spin Echo

- **Advantages**
  - Turbo factor accelerates imaging
  - Can be used with 2D slice interleaving
  - Allows T_2 weighted imaging in a breath hold
- **Disadvantages**
  - High turbo factors (ETL>4):
    - Blur images
    - Alter image contrast
  - Fat & Water are both bright on T_2-weighted
    - Water/CSF T_2 is long
    - Repeated 180s reduce spin-spin interaction
      - This lengthens the moderate T_2 of fat
  - SAR can be high

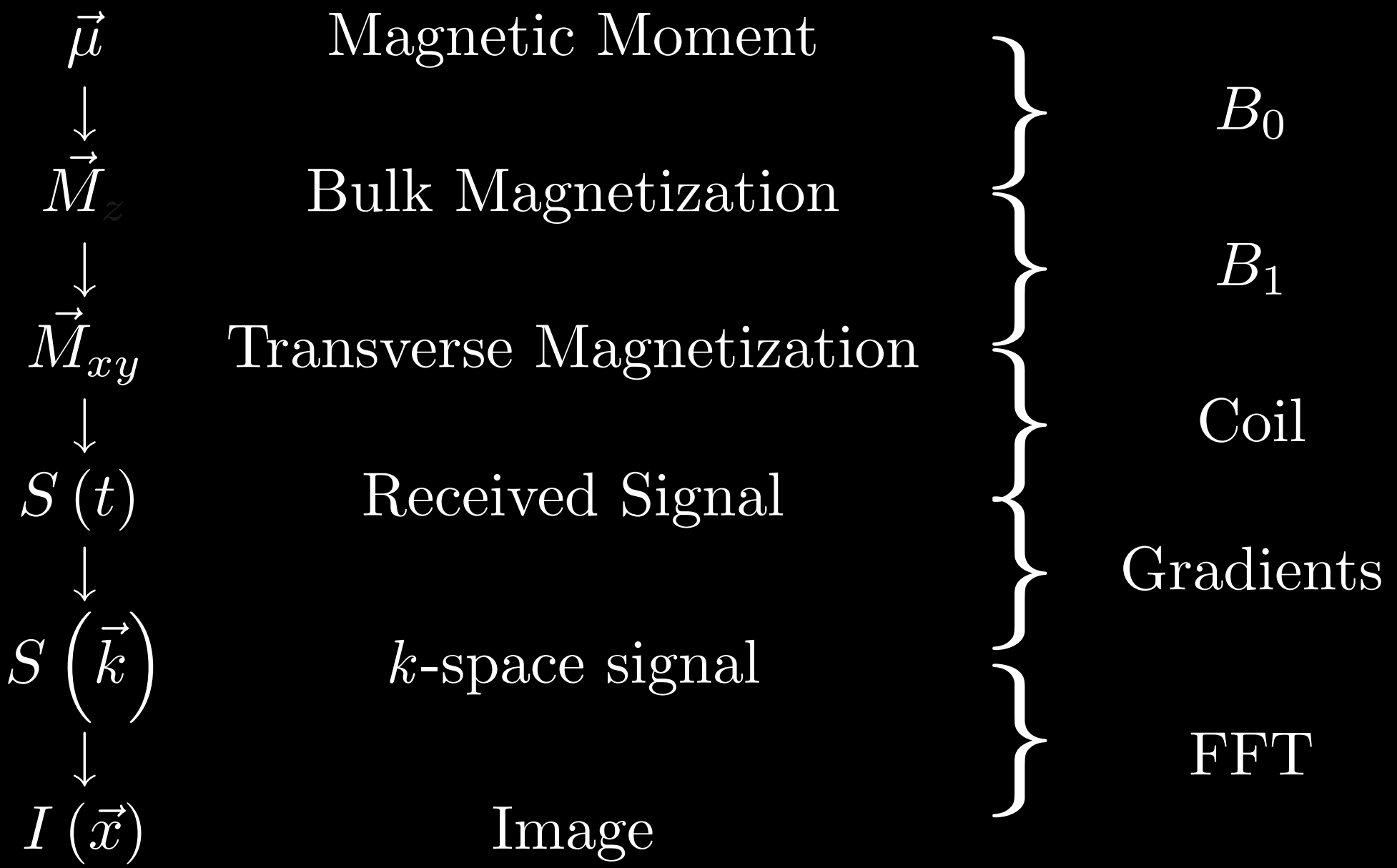
Spin Echo EPI



Spin Echo EPI

- **Advantages**
  - Can acquire data in a “single shot”
  - Can be used with 2D slice interleaving
  - Allows T_2^* weighted imaging in a breath hold
- **Disadvantages**
  - Single Shot EPI
    - Ghosting
    - Blur images
    - Image distortion
    - Alter image contrast
  - Multi-shot EPI
    - Slower than single shot
      - Faster than SE
- **Applications**
  - DWI, Perfusion, fMRI

Dipoles to Images



Some consolation...

Thanks

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