Magnetic Resonance Imaging

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The Plan

- The Magnetic Resonance Phenomenon & Contrast (30)
- Spatial Encoding (26)
- The "Pulse Sequence" Rules Everything (3)

Seventh Inning Stretch

- Fast Imaging (14)
- Functional MRI (18)
- Diffusion and Summary (9)

- Image Quality and Artifacts (48)

Metaphor

|'metəˌfôr; -fər|
noun

1. a figure of speech in which a word or phrase is applied to an object or action to which it is not literally applicable :

"I had fallen through a trapdoor of depression," said Mark, who was fond of theatrical metaphors | her poetry depends on suggestion and metaphor.

a thing regarded as representative or symbolic of something else, esp. something abstract :

the amounts of money being lost by the company were enough to make it a ***metaphor for*** *an industry that was teetering.*

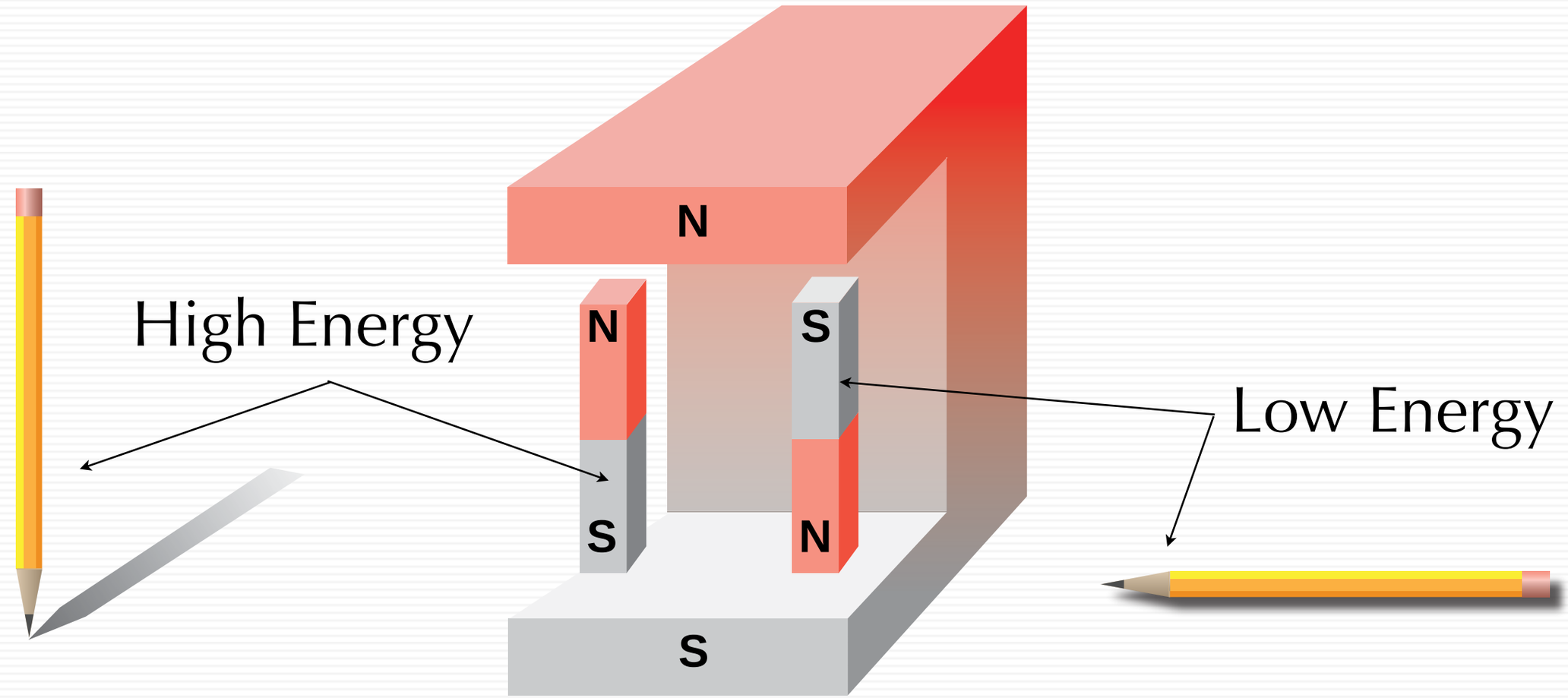
2. Little white lie

Spin

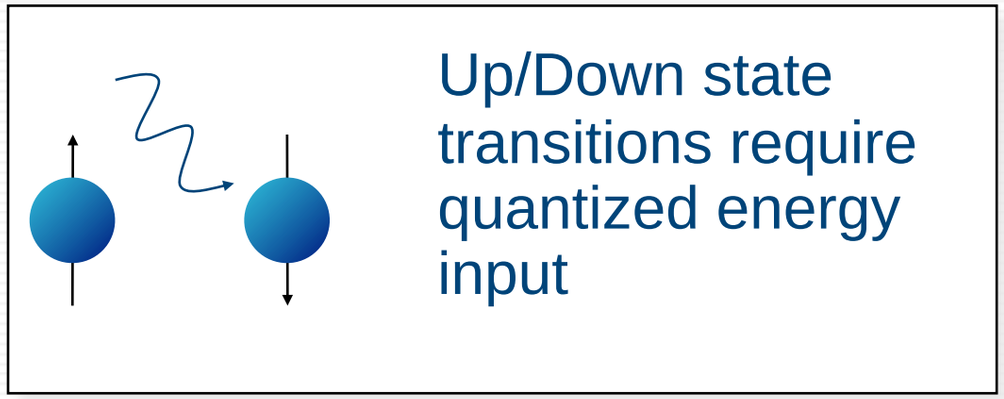
- Similar to Angular Momentum
- Intrinsic Property of Matter (rotation *not* required)
- Spin takes half-integer (0, 1/2, 1, 1-1/2) quantum values
  - Particles with integer spin are ***Bosons***
  - Particles with half integer spins are ***Fermions***
- Atoms with odd Mass Numbers (^{1}H, ^{13}C, ^{19}F, ^{23}Na) have half-integer spin.
- Electrons, Neutrons, Protons have spin = 1/2

Spin States

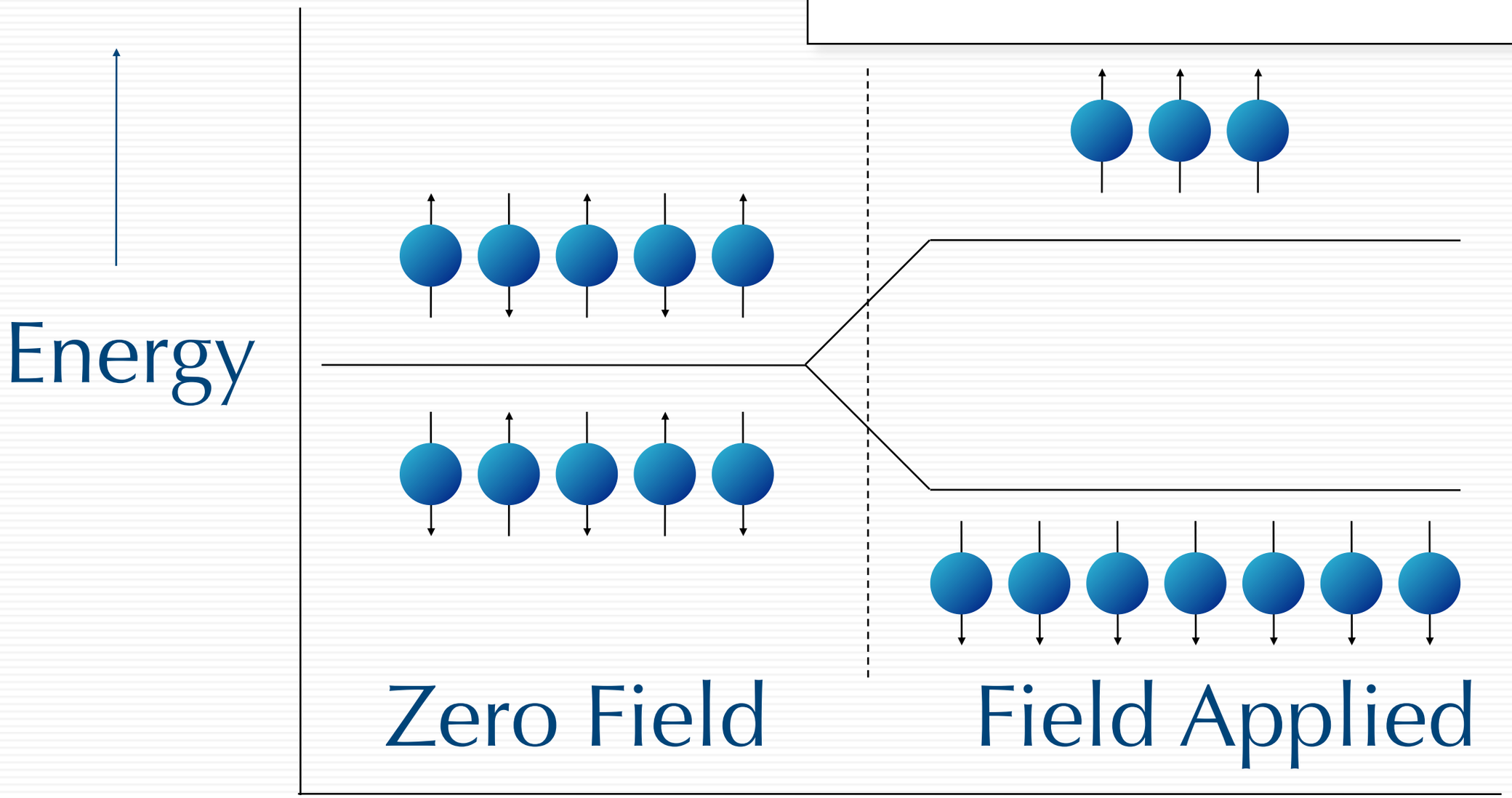
- A Spin 1/2 particle has two states: up and down
- In a magnetic field, B_0, the two states have different energies



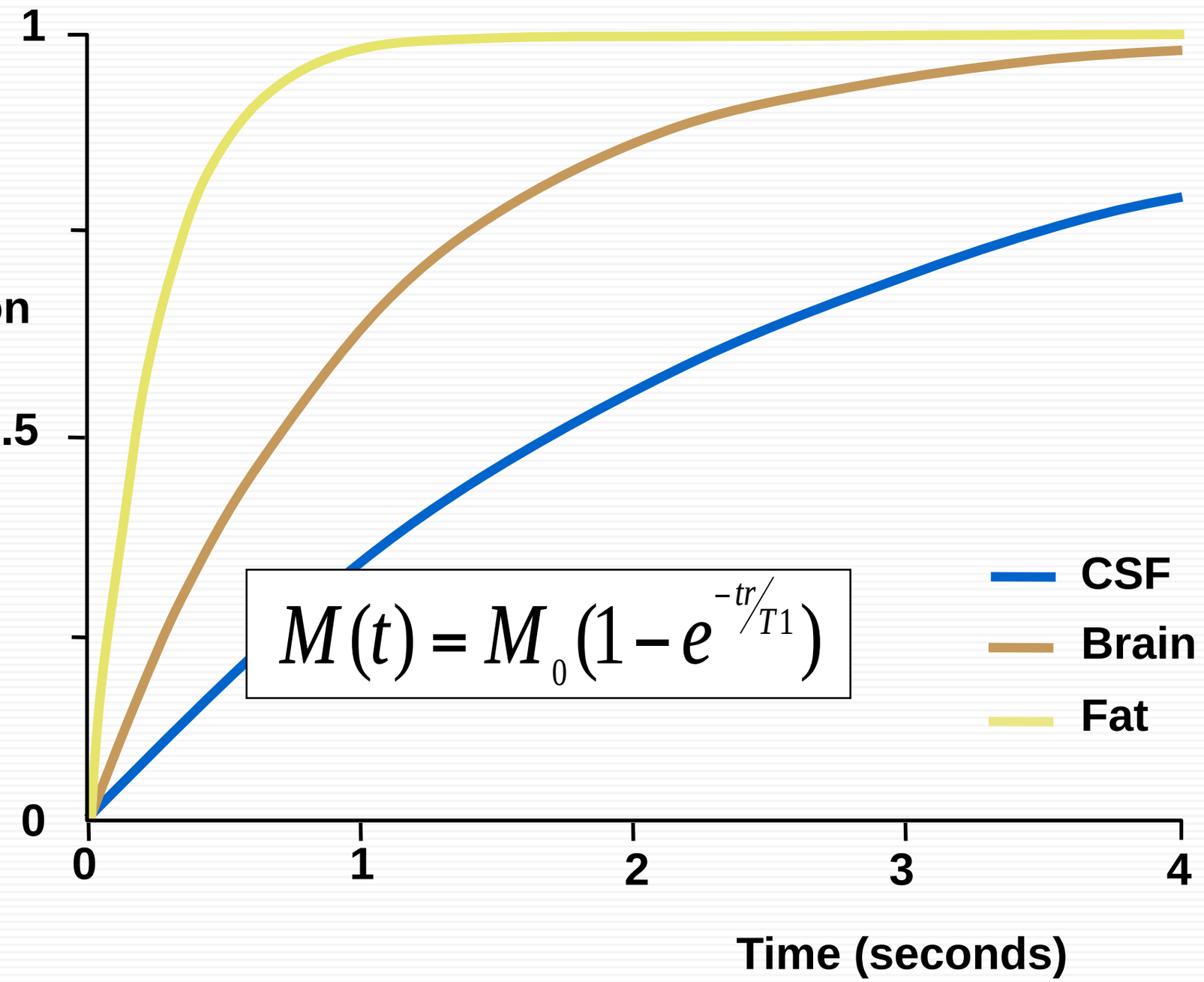
Transition to Equilibrium



Up/Down state transitions require quantized energy input

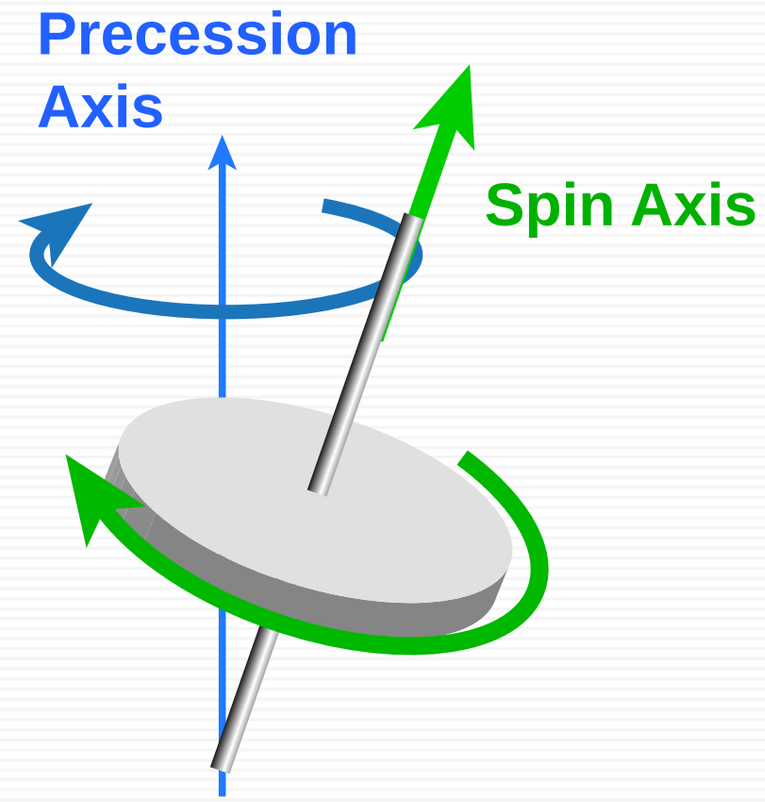


T1



Nuclear Spin

- Angular momentum is a vector quantity having *magnitude & direction*.
- Angular Momentum is *Conserved*, causing precession to occur



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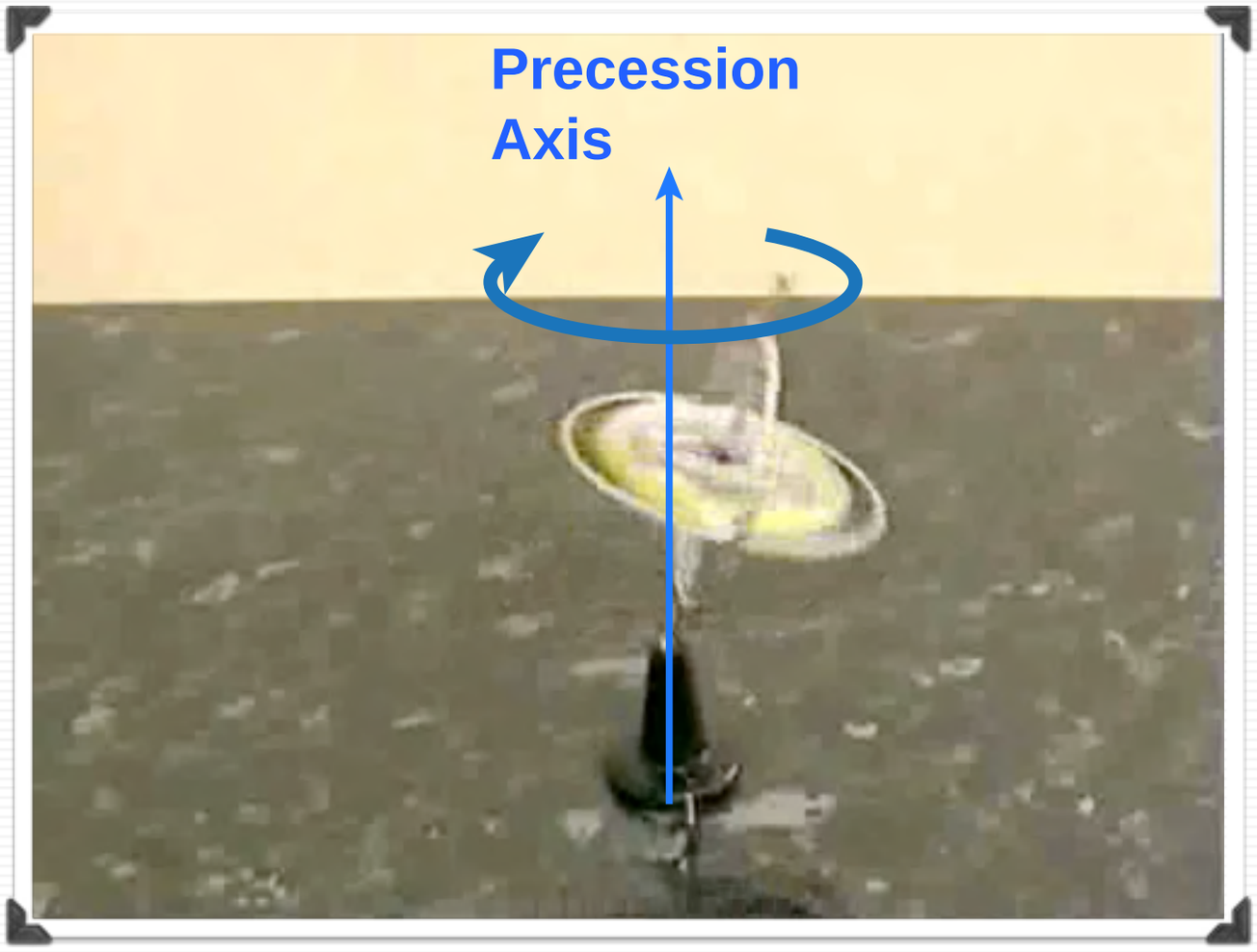
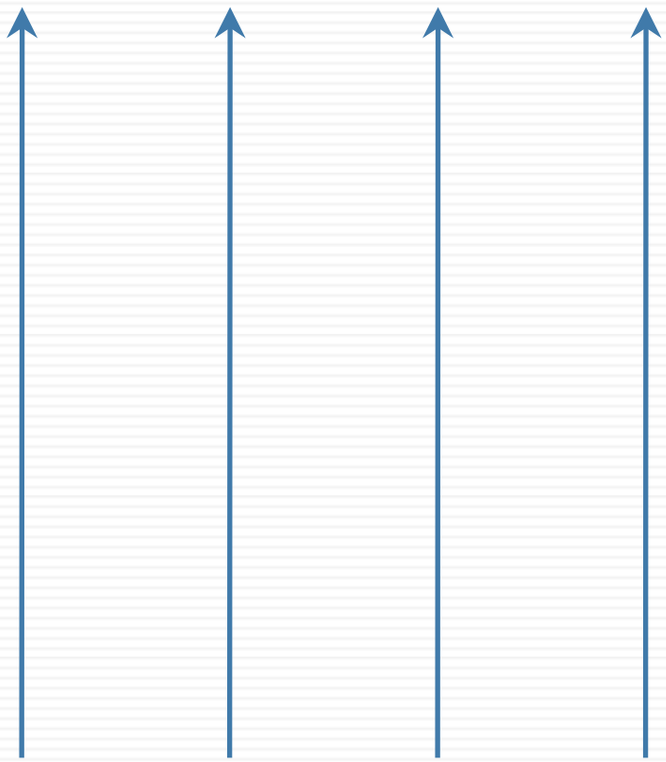
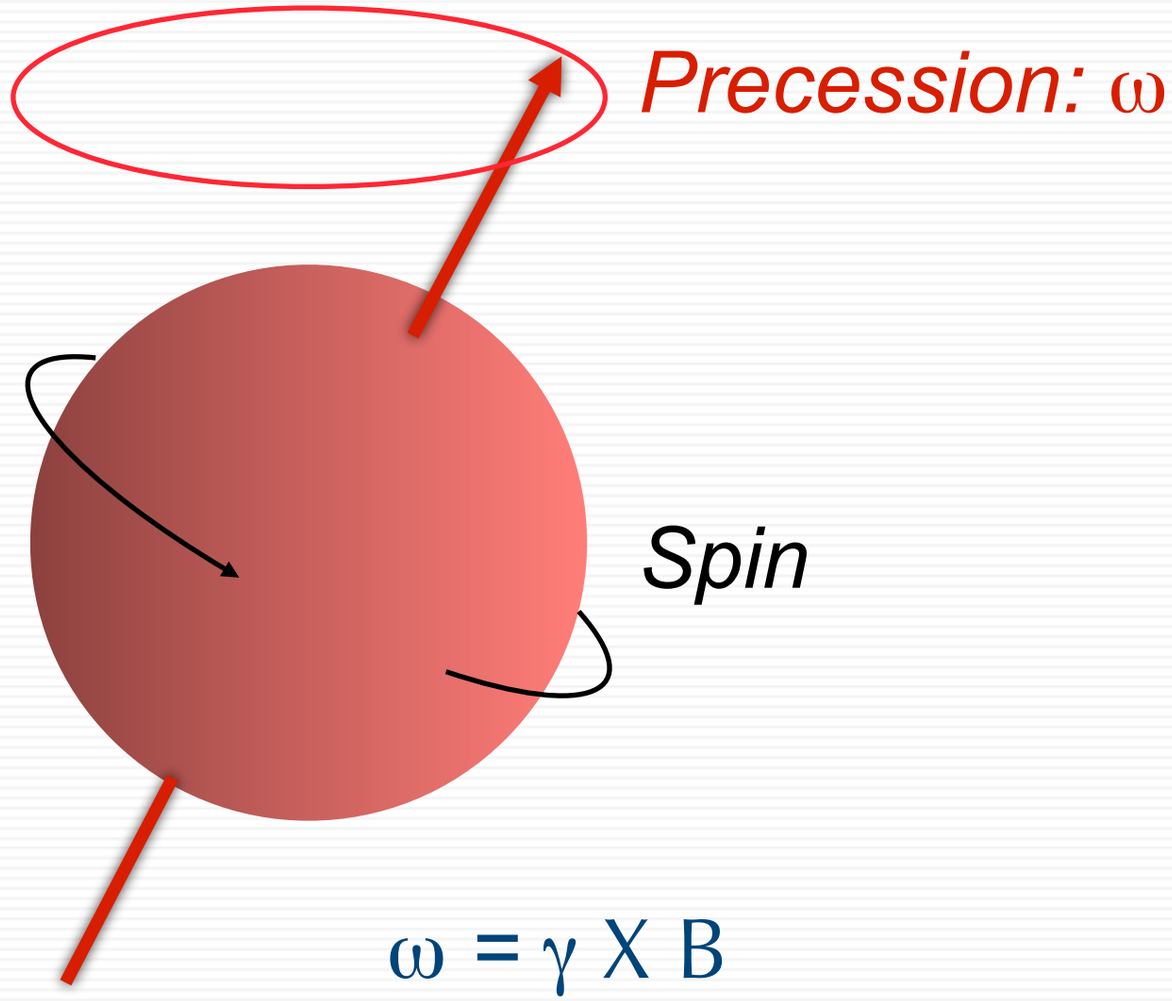


Proton Precession

Due to their angular momentum, Protons precess in the magnetic field.



Applied Magnetic Field: **B**



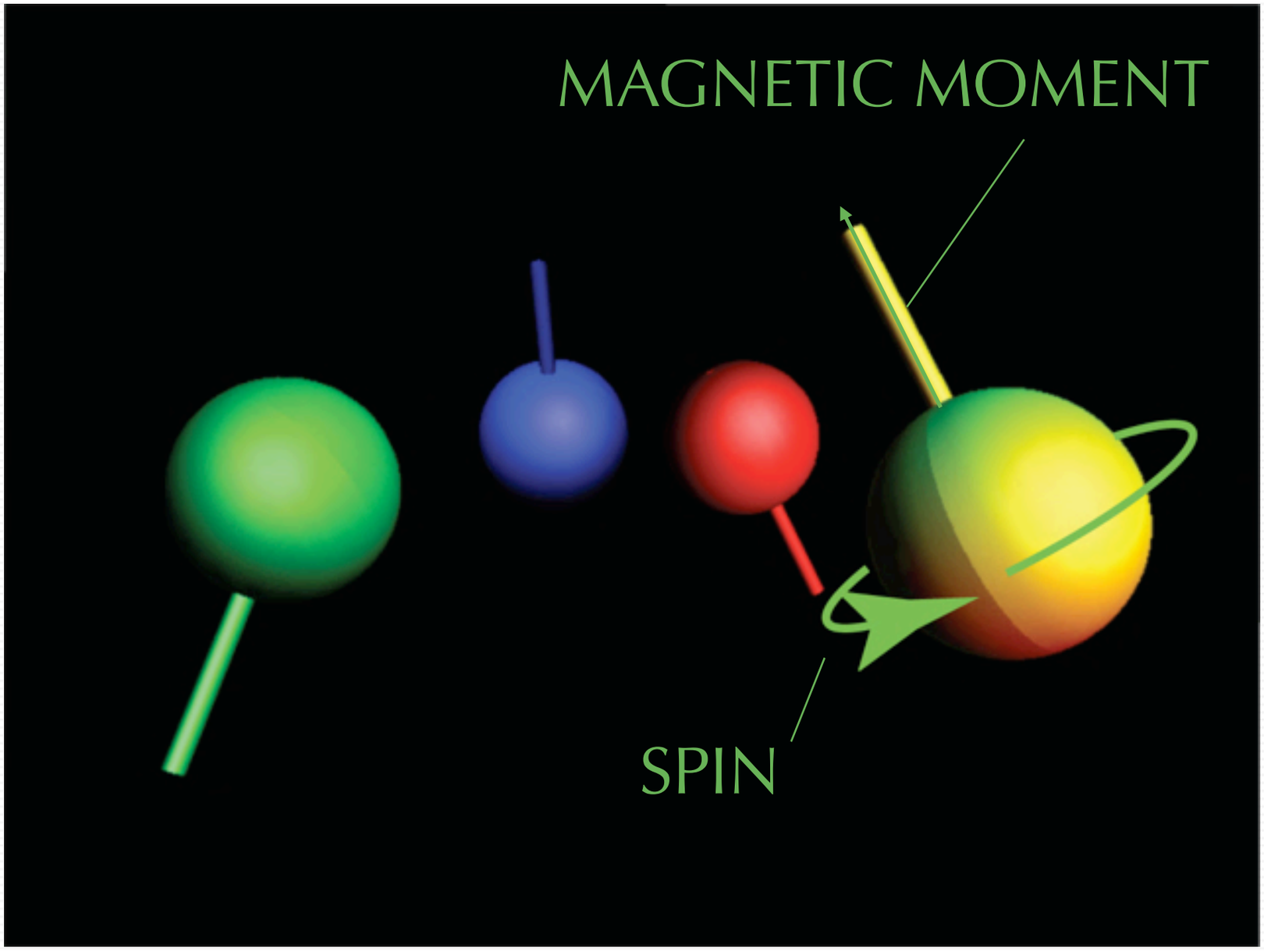
$$\omega = \gamma \times B$$

$$\gamma_H \approx 267.52 \text{ Rad/sec/Tesla}$$

$$\approx 42.577 \text{ MHz/Tesla}$$

MRI 101a

Protons, the Nucleus of Hydrogen, Have a Magnetic Moment



MRI 101b

Protons Align (polarize) and Precess in Magnetic Fields

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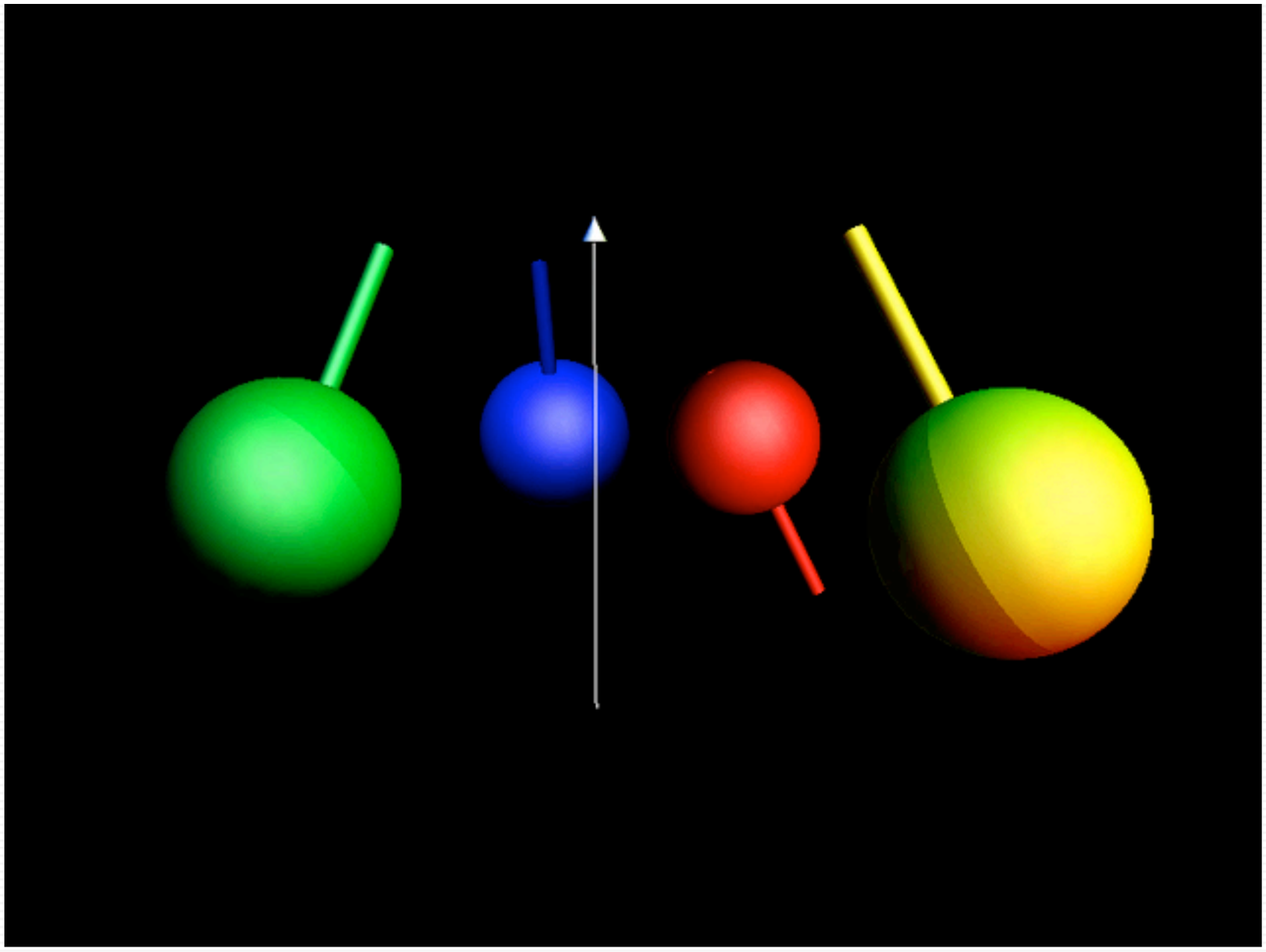


Proton Responses to Magnetic Field

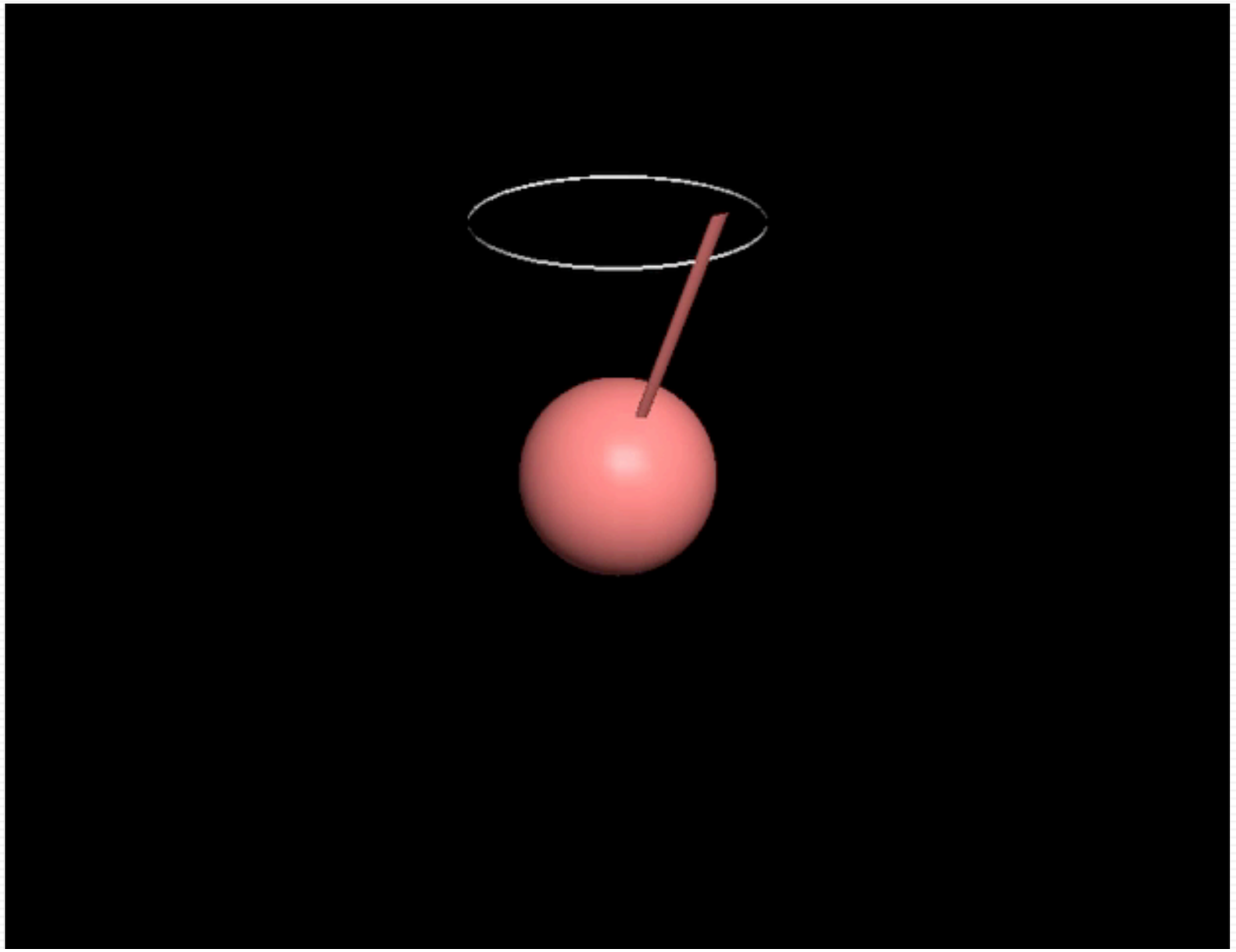
- Spin Alignment Along Net Applied Field
  spins align parallel or anti-parallel to the applied field

- Precession About the Magnetic Field
  at a precession frequency of: $\gamma \times B$, known as the Larmor frequency

- Spin Alignment Occurs at the Rate, T1

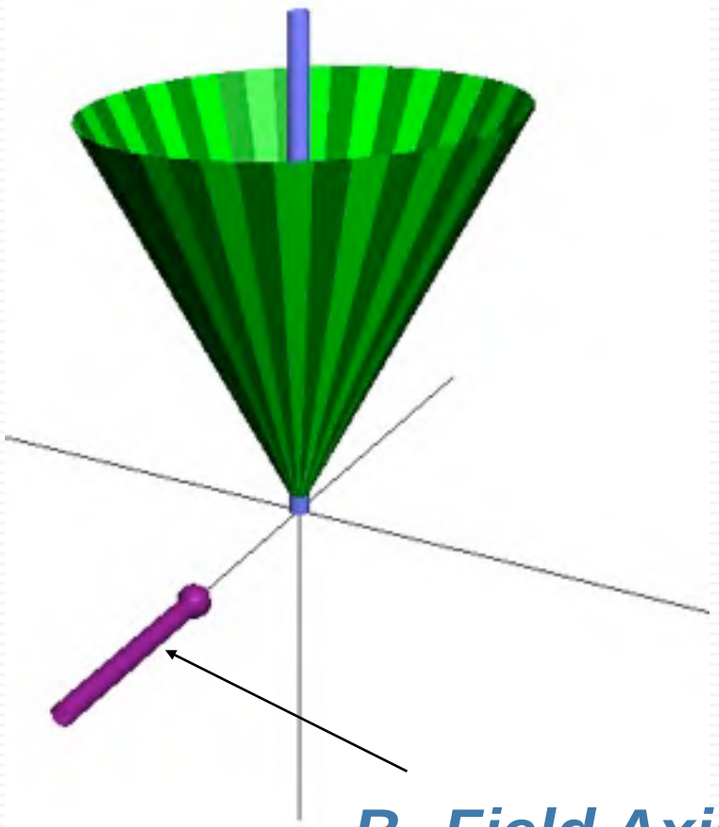
Vector Addition of Spins

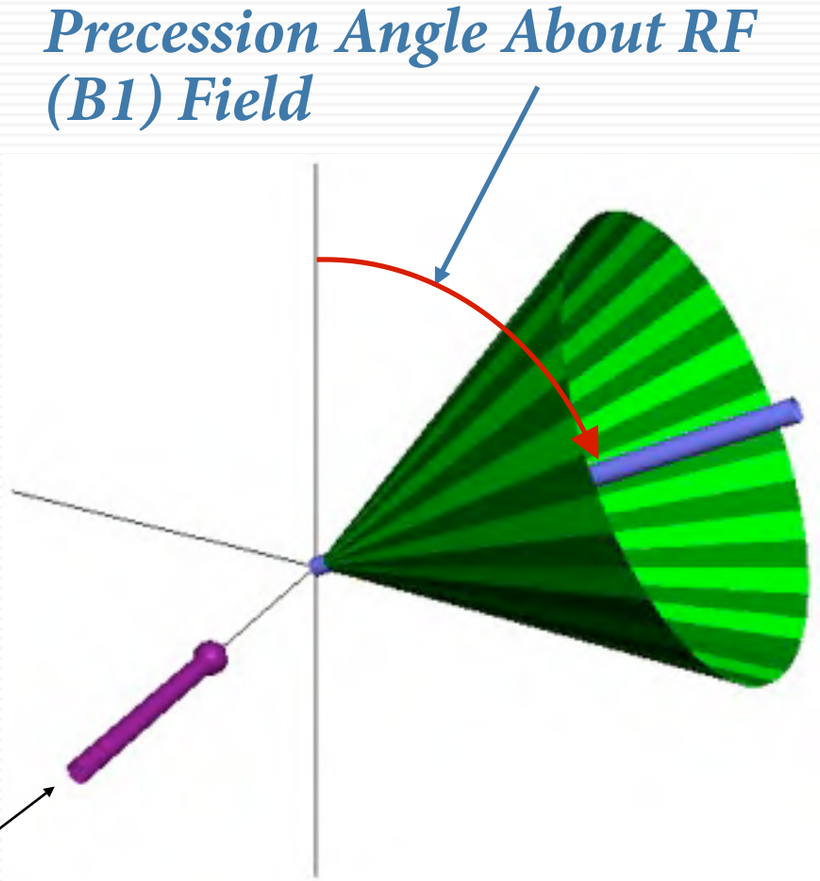
Vector Addition of Spins



the Resonance Phenomenon

Static Magnetic Field:
B_0

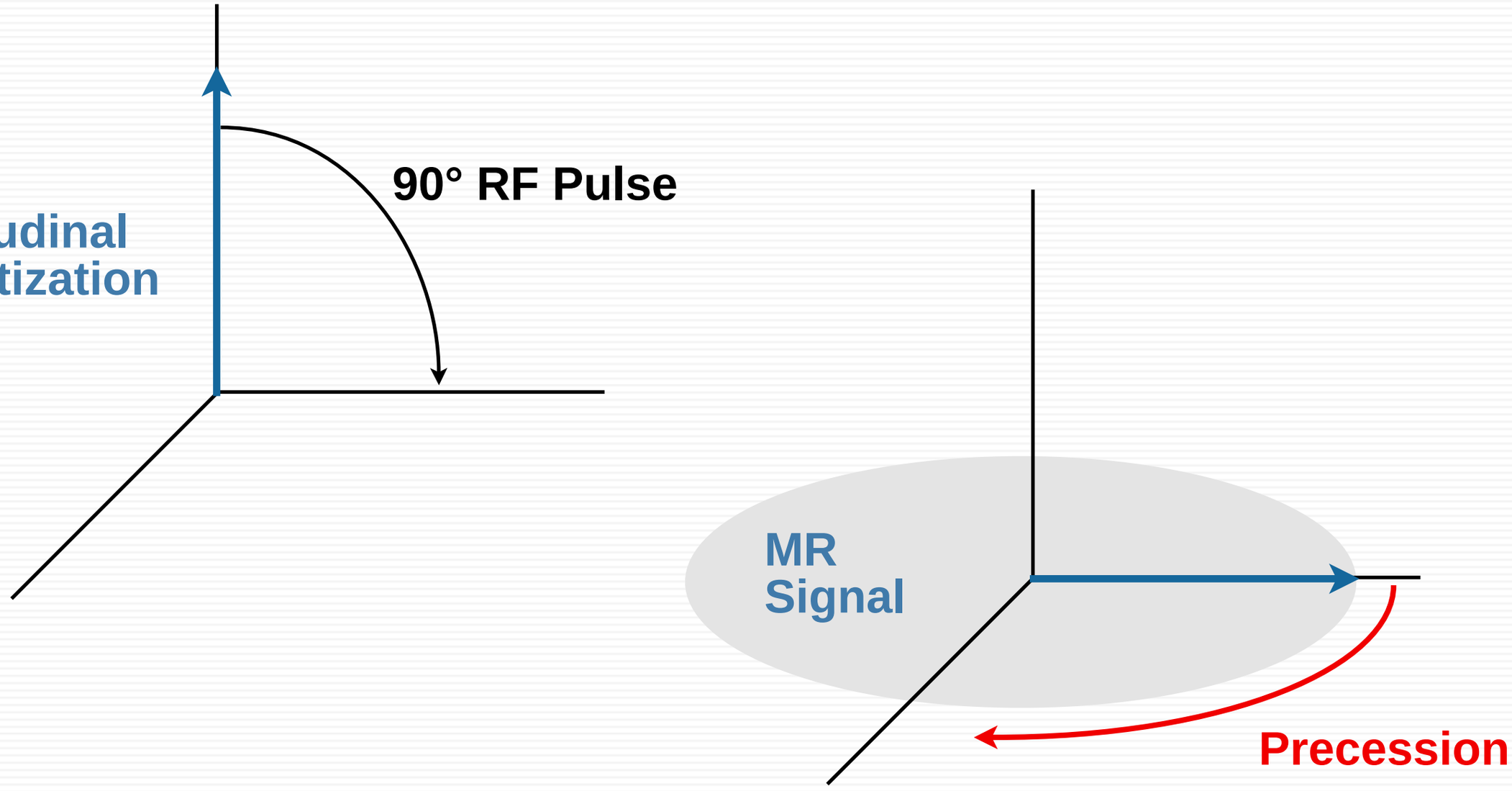




B_1 Field Axis

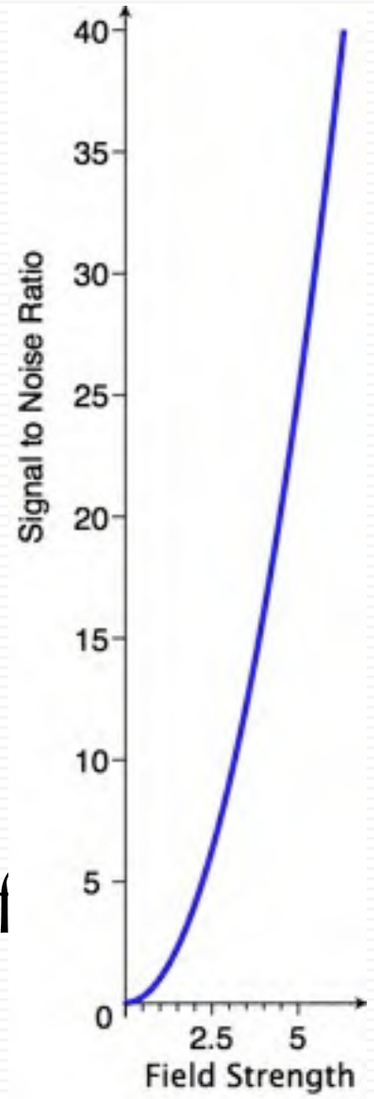
- When a second magnetic field (B_1) is applied, rotating at the Larmor rate, the proton will precess about it.
- The duration and amplitude of B_1 determine rotation angle

RF Pulses Convert Longitudinal Magnetization to MR Signal



Inductive Detection

With Conventional MRI, the Signal is Detected Inductively.
Therefore the Intensity is Proportional to Frequency



In This Mode, Signal to Noise Ratio Falls off Rapidly With Field Strength

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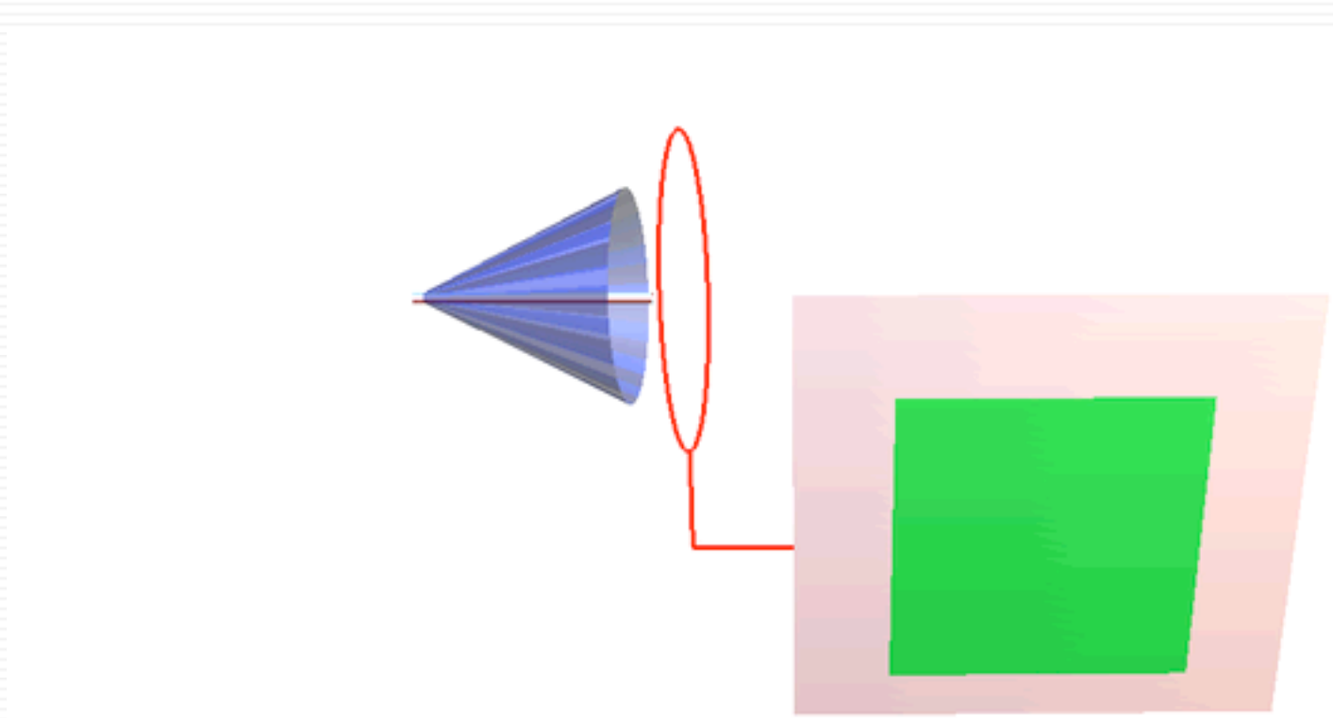

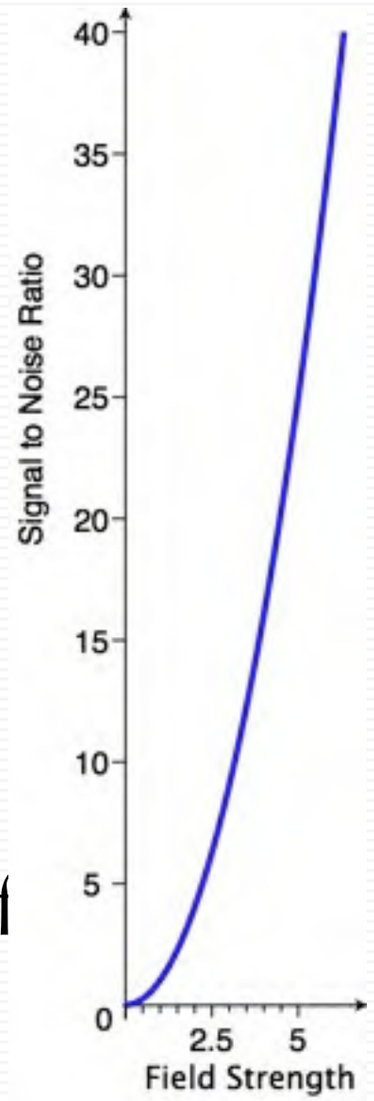

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From Magnetization to Signal

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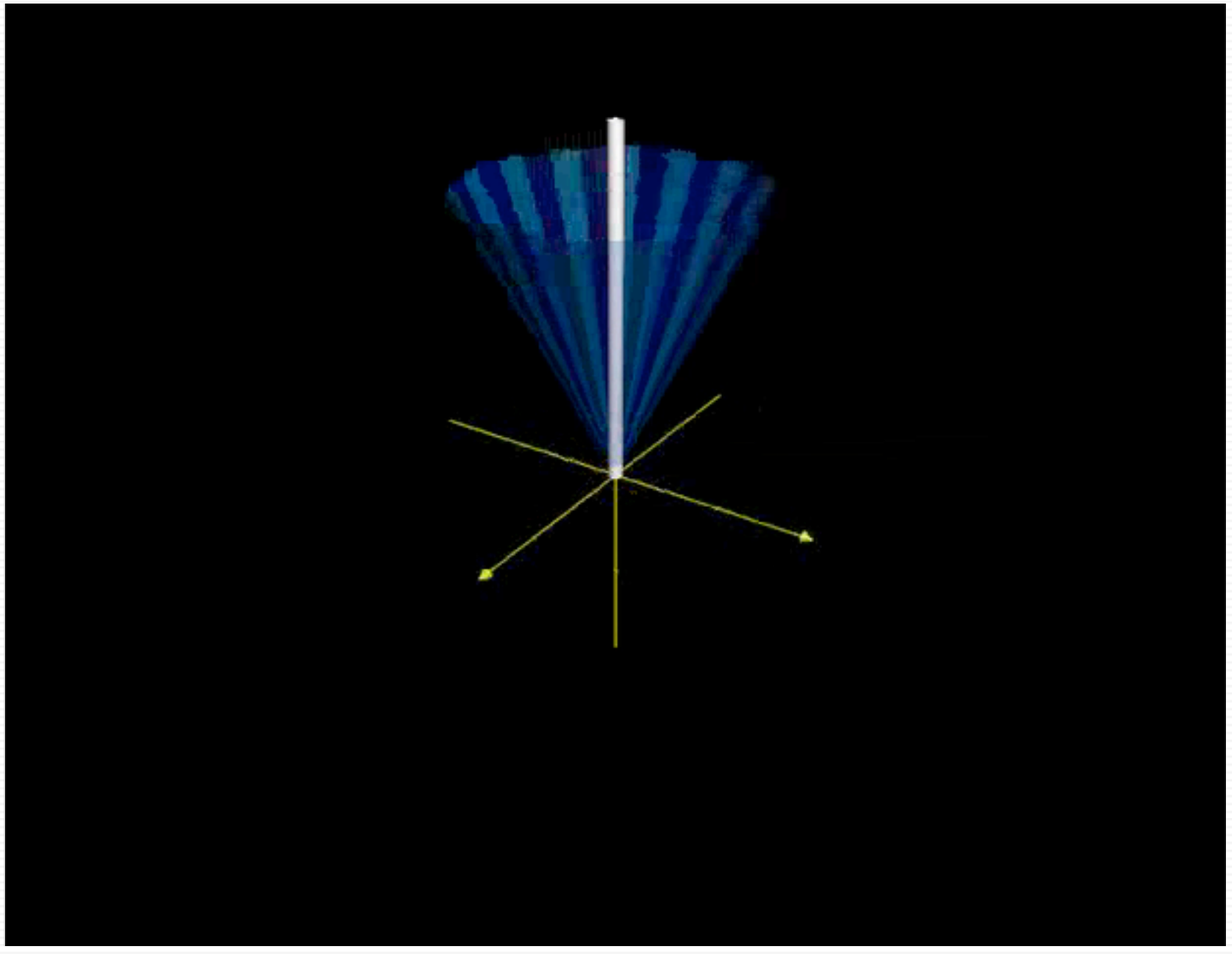
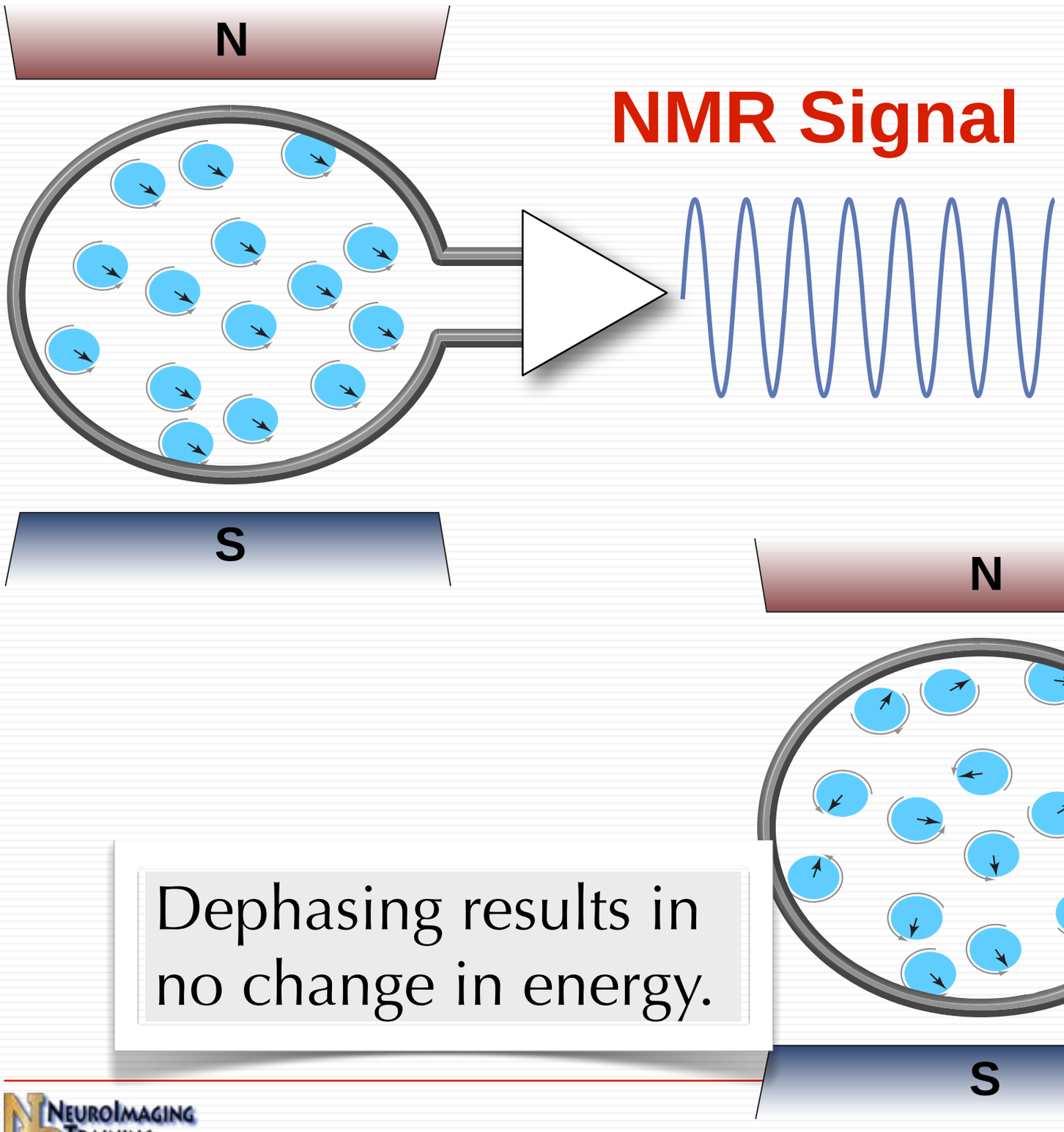
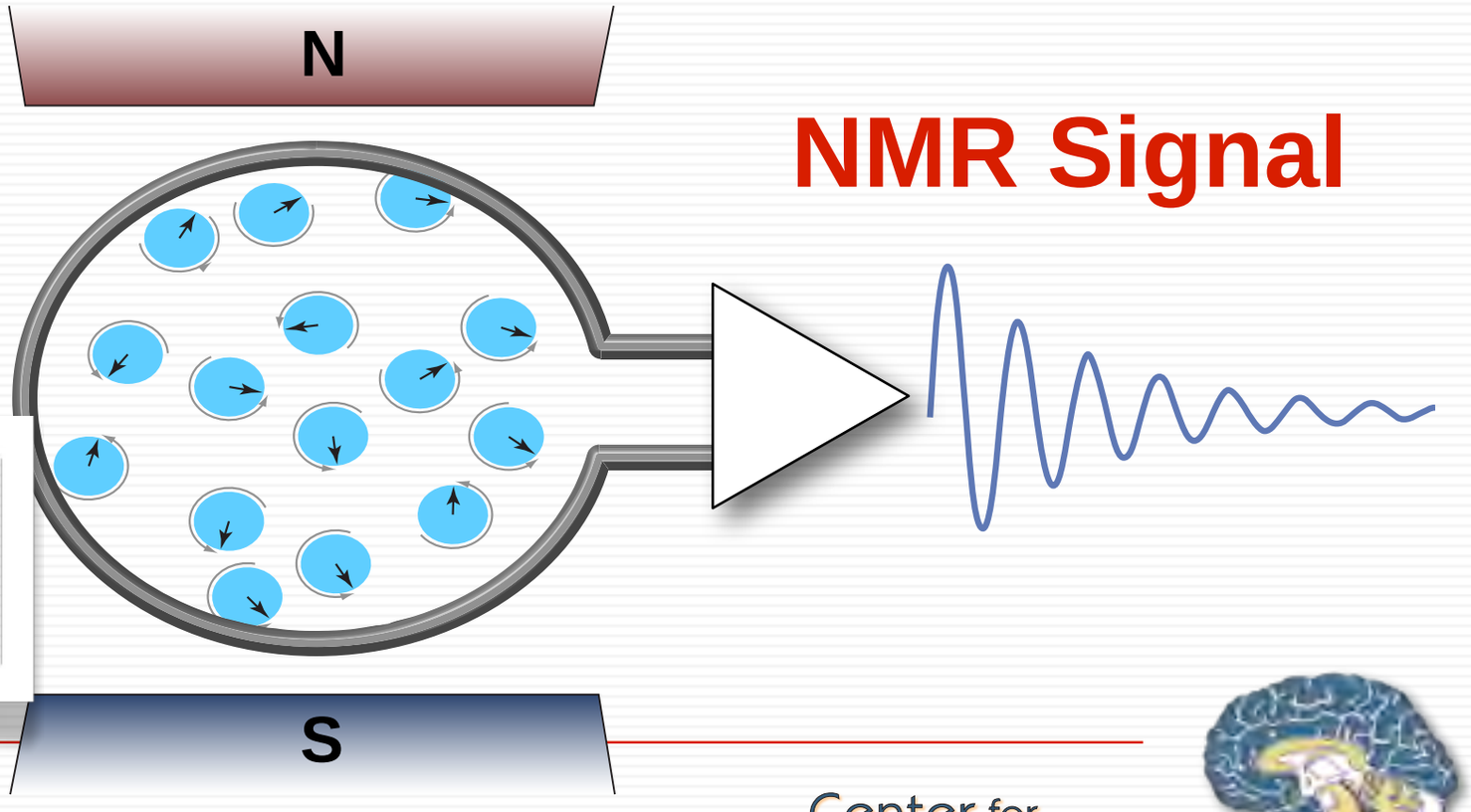


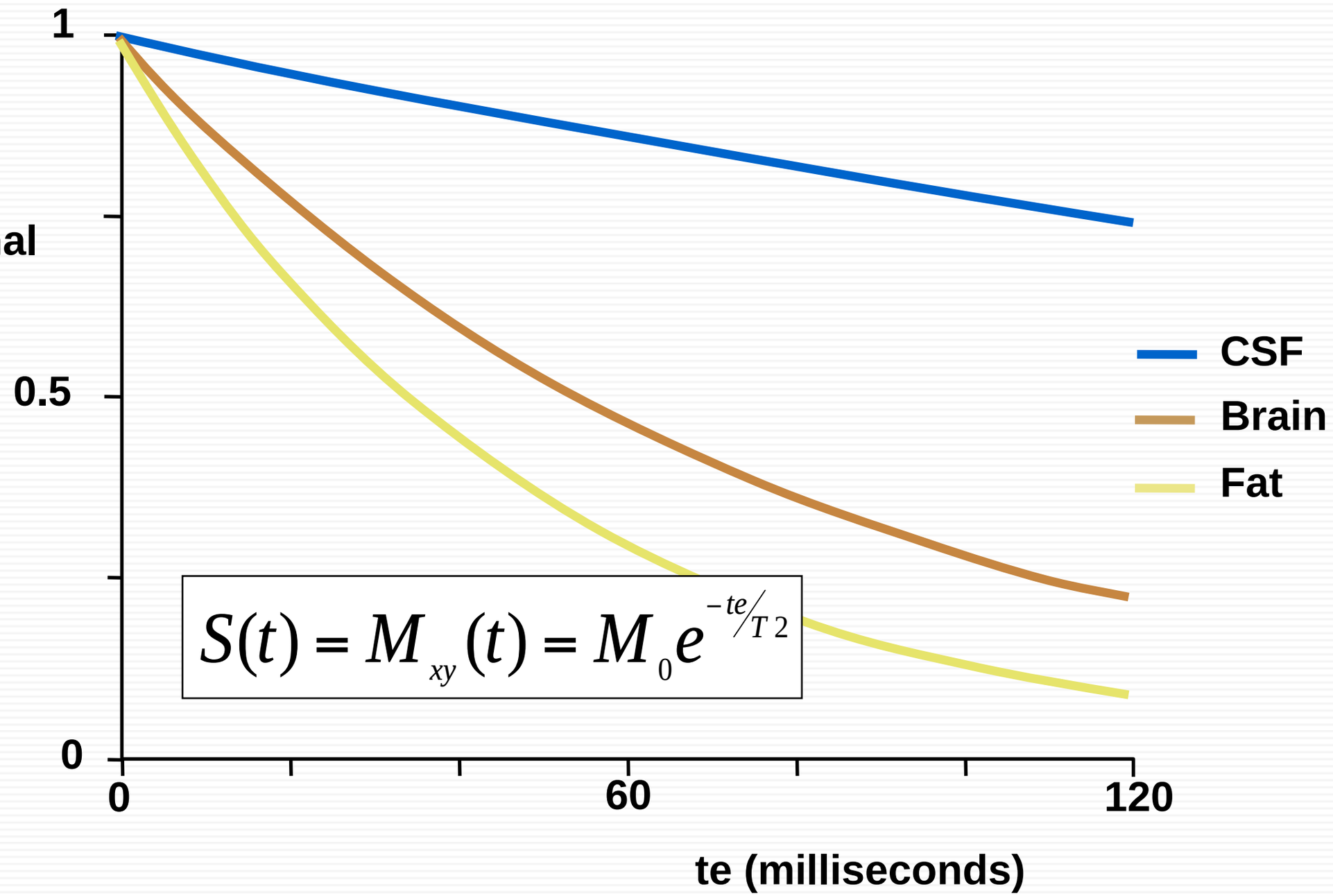
Spin Dephasing



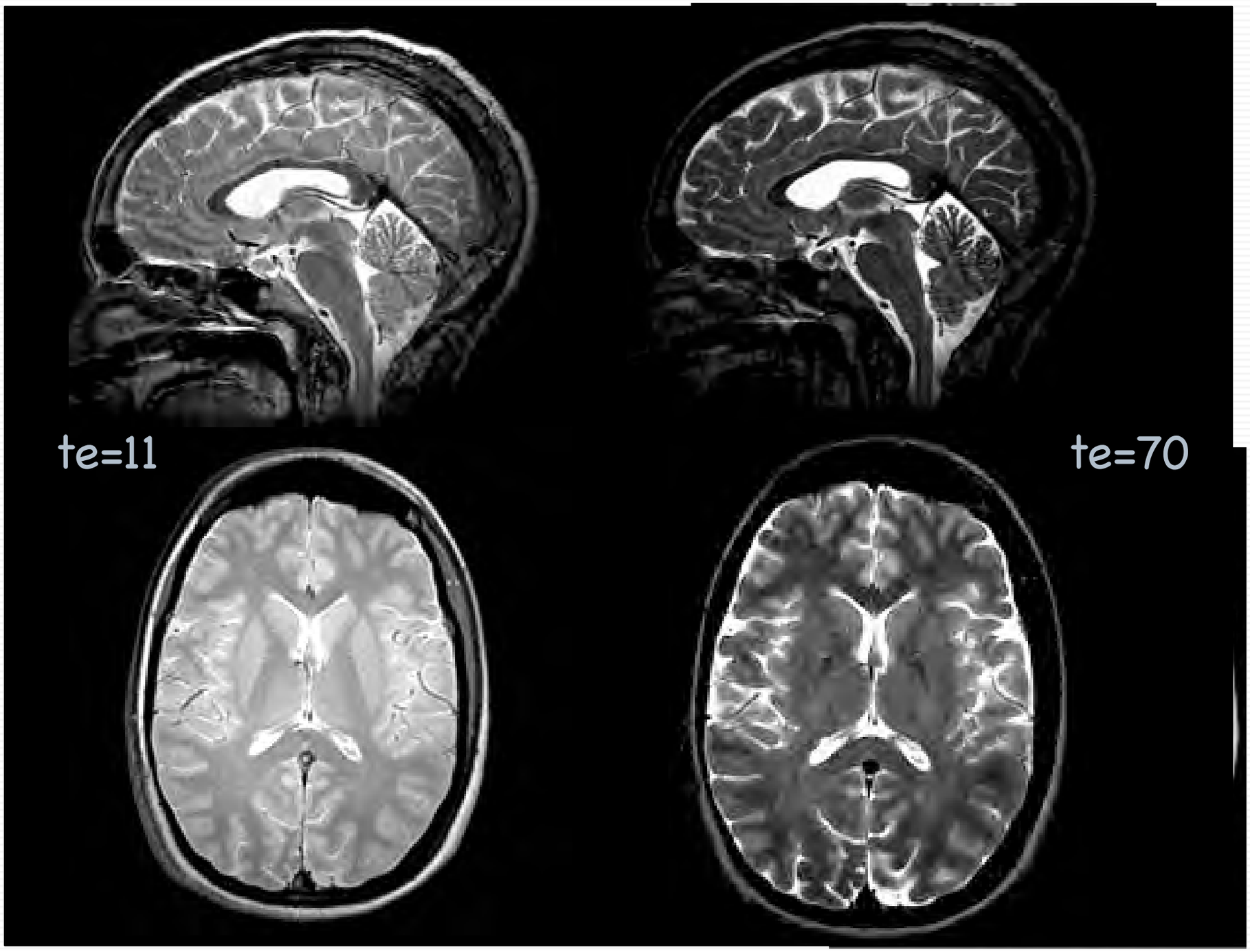


Dephasing results in no change in energy.

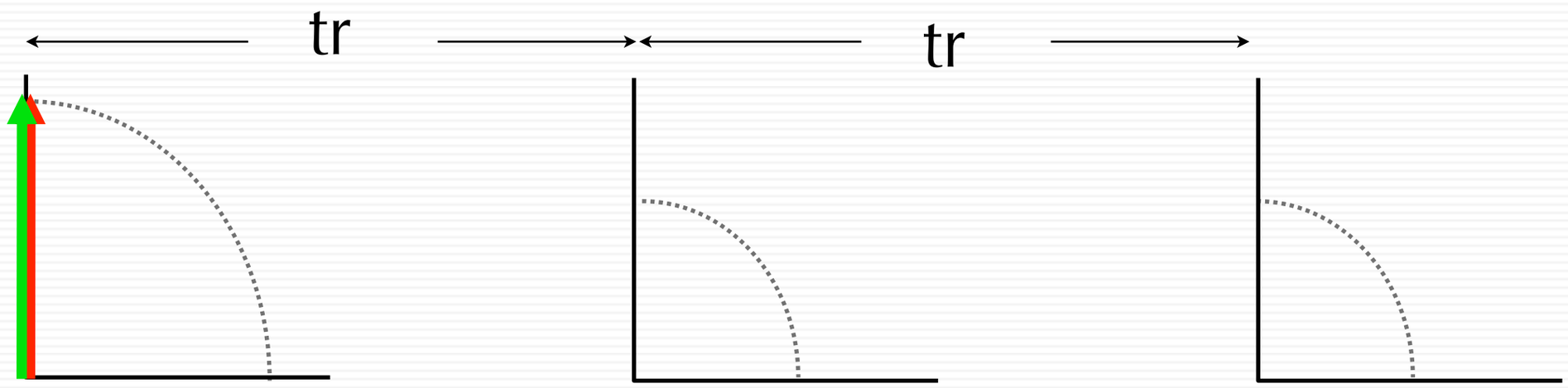
T2 and te



Effects of TE at long TR



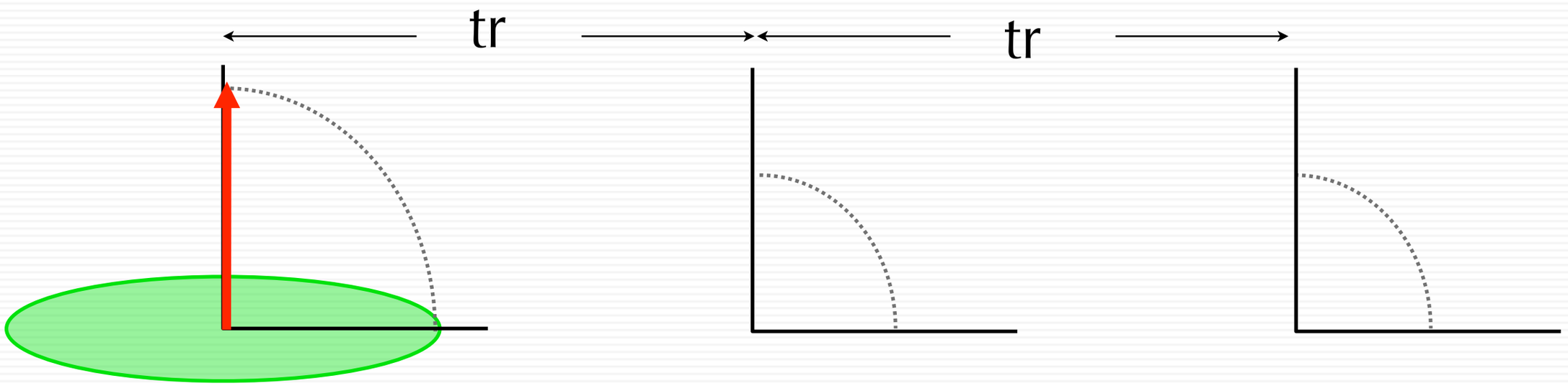
Partial Saturation Sequence



Sequence of 90° Pulses

NMR Signal

Partial Saturation Sequence



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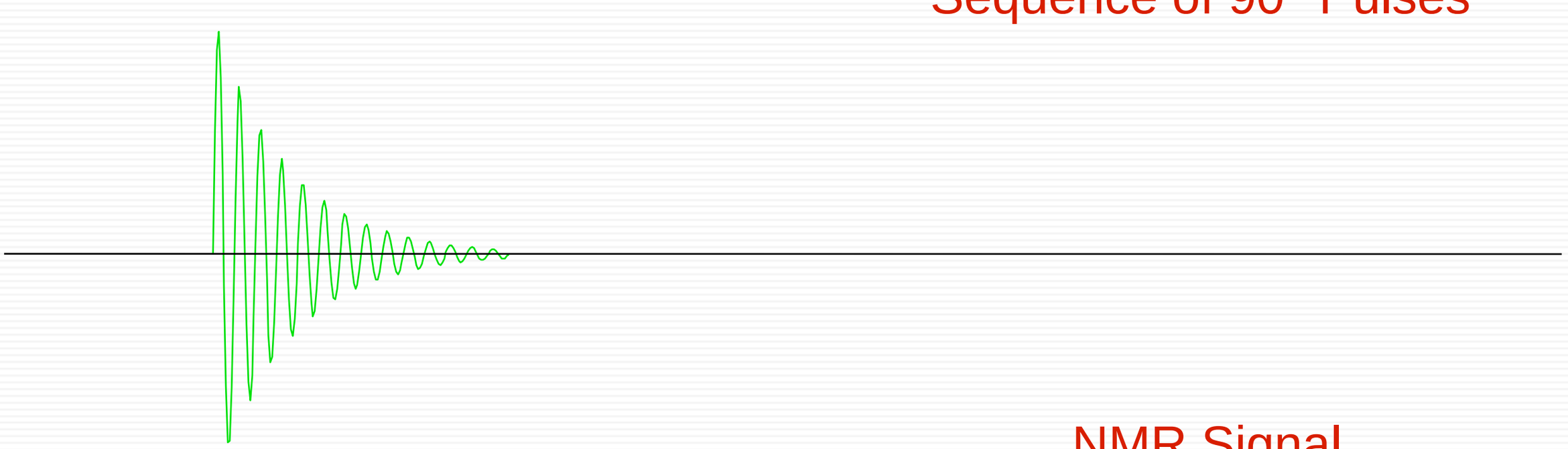

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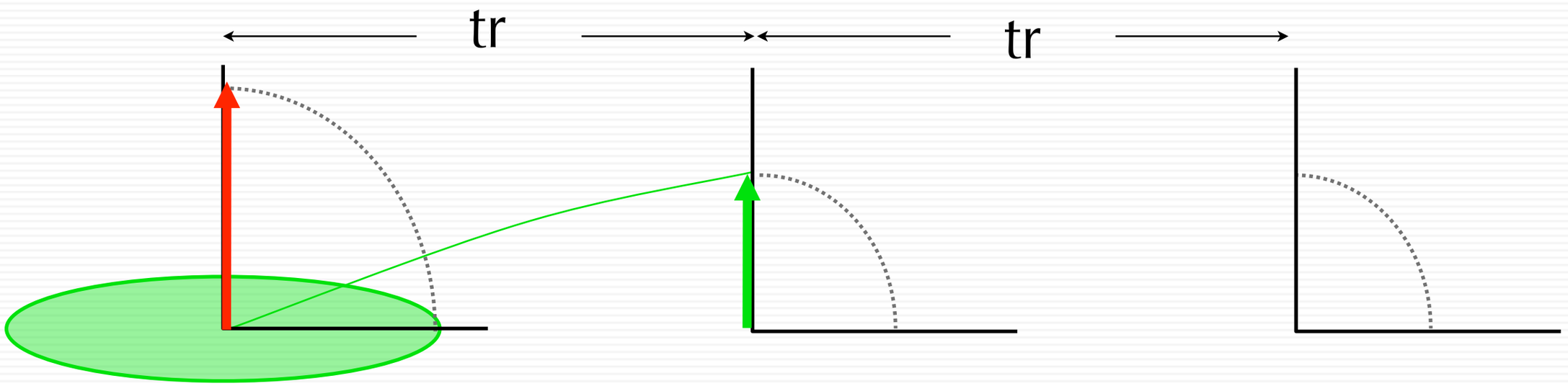

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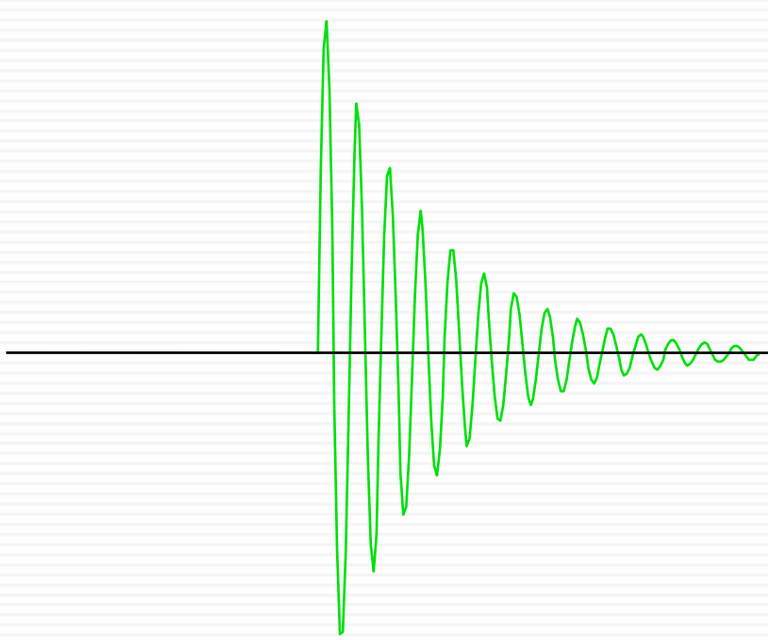

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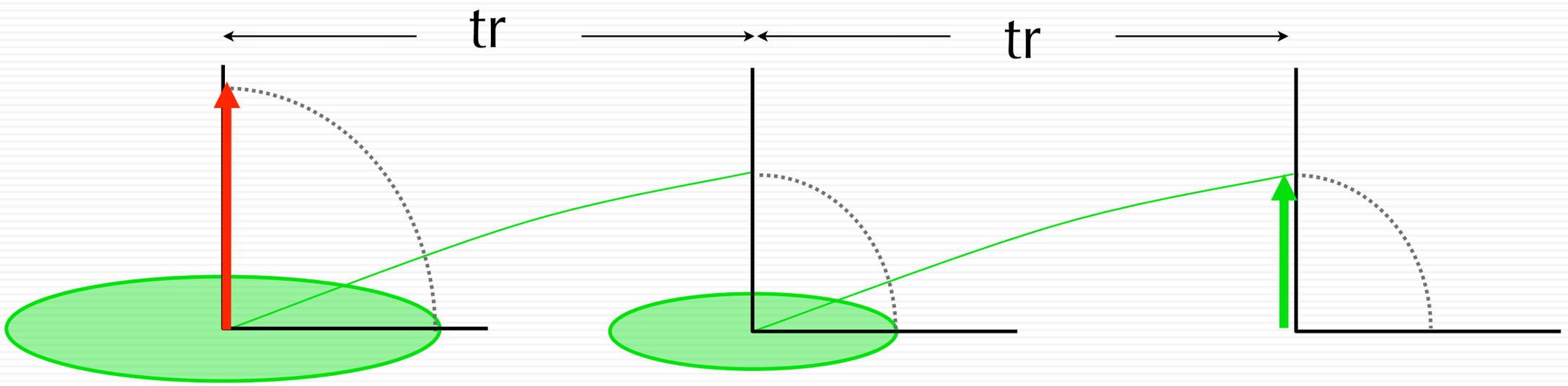

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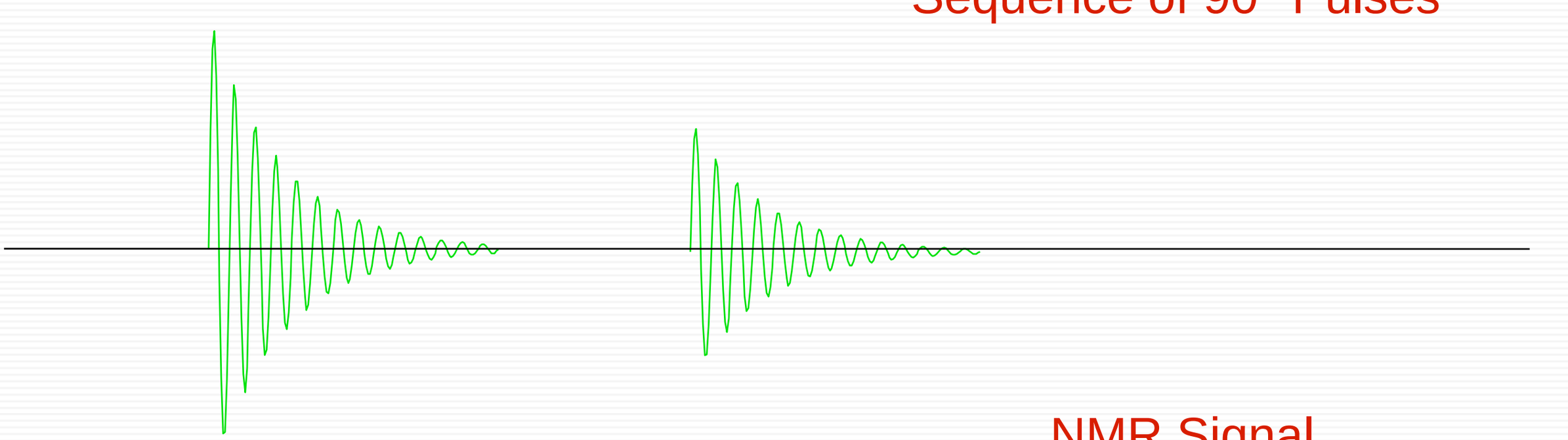


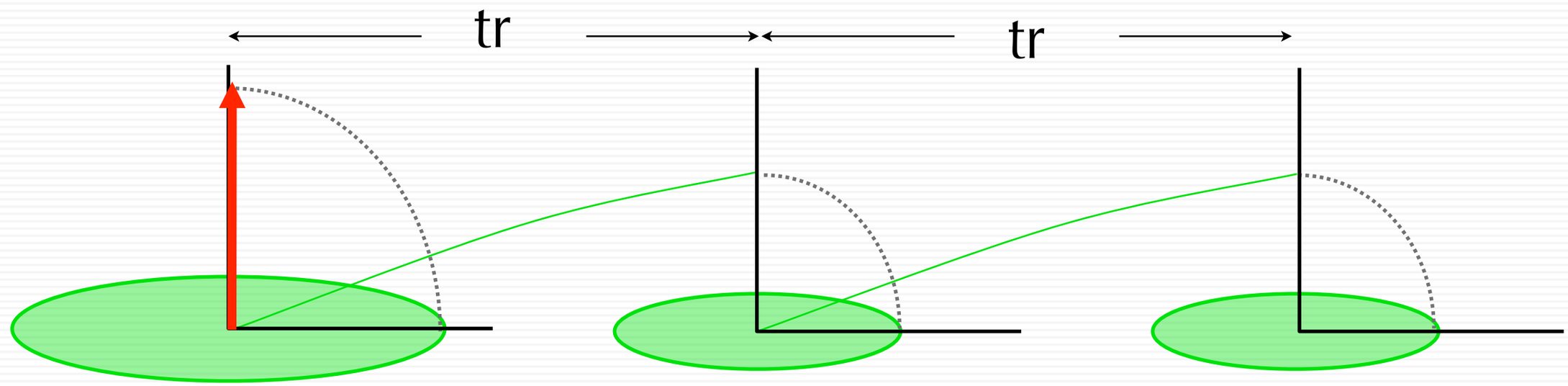
NMR Signal

NeuroImaging Training Program

www.brainmapping.org

Center for Cognitive Neuroscience

Partial Saturation Sequence



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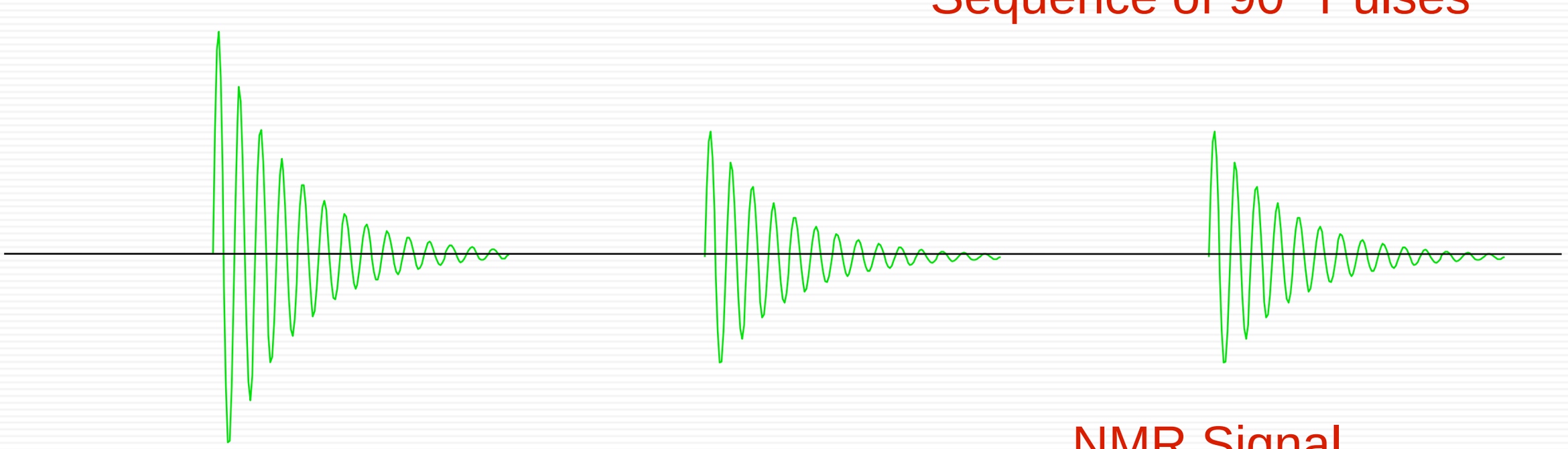

NMR Signal

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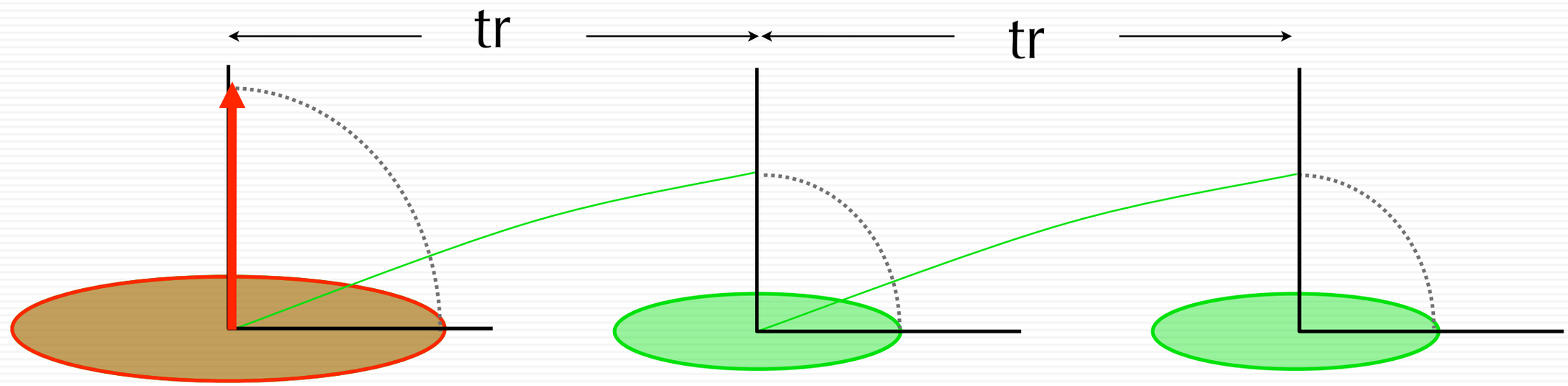
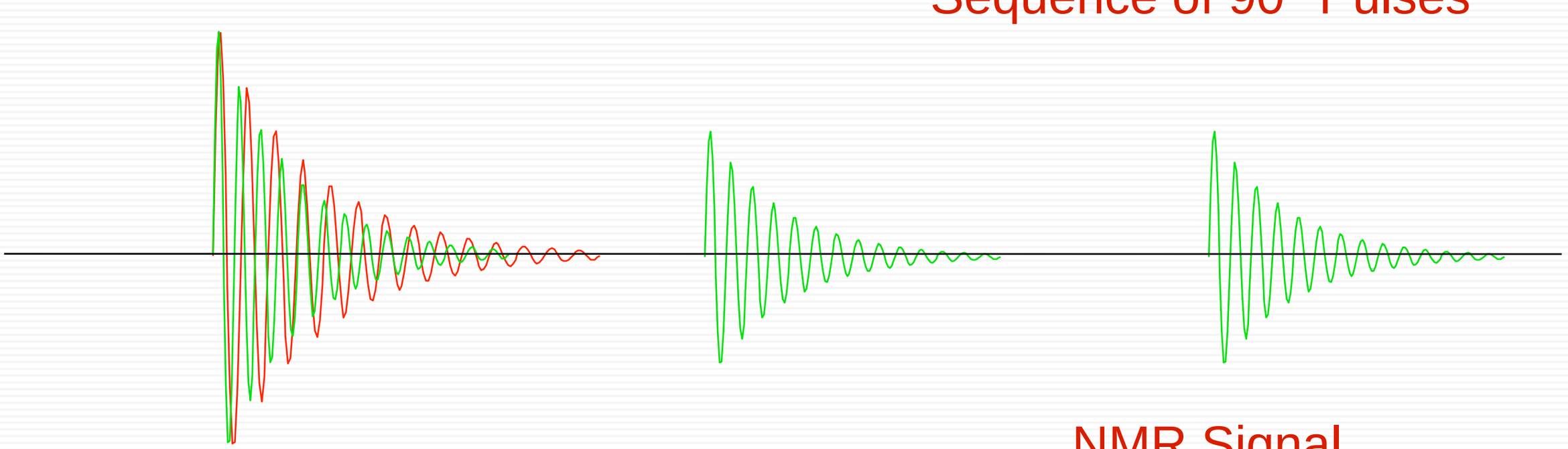


Sequence of 90° Pulses



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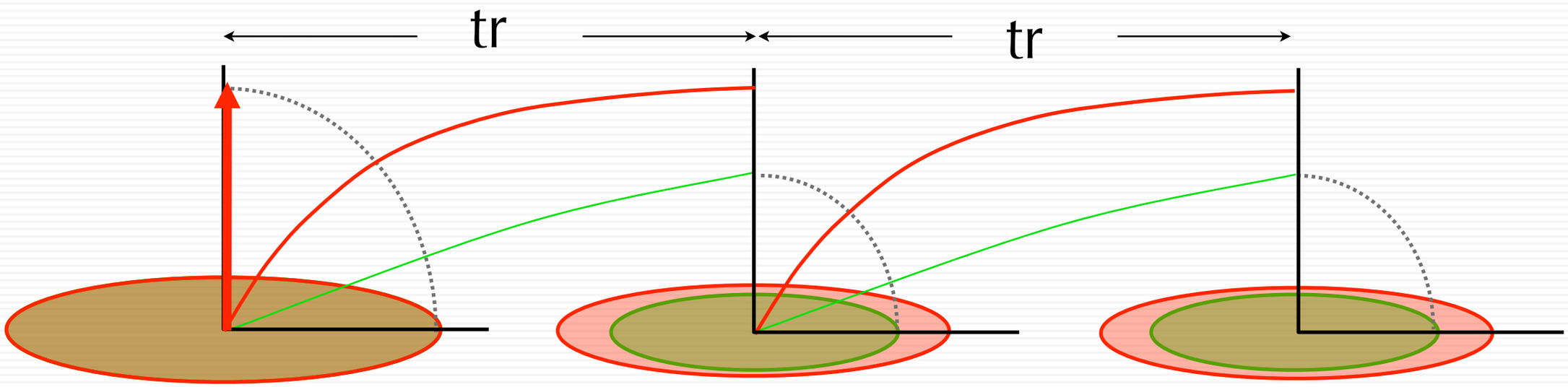

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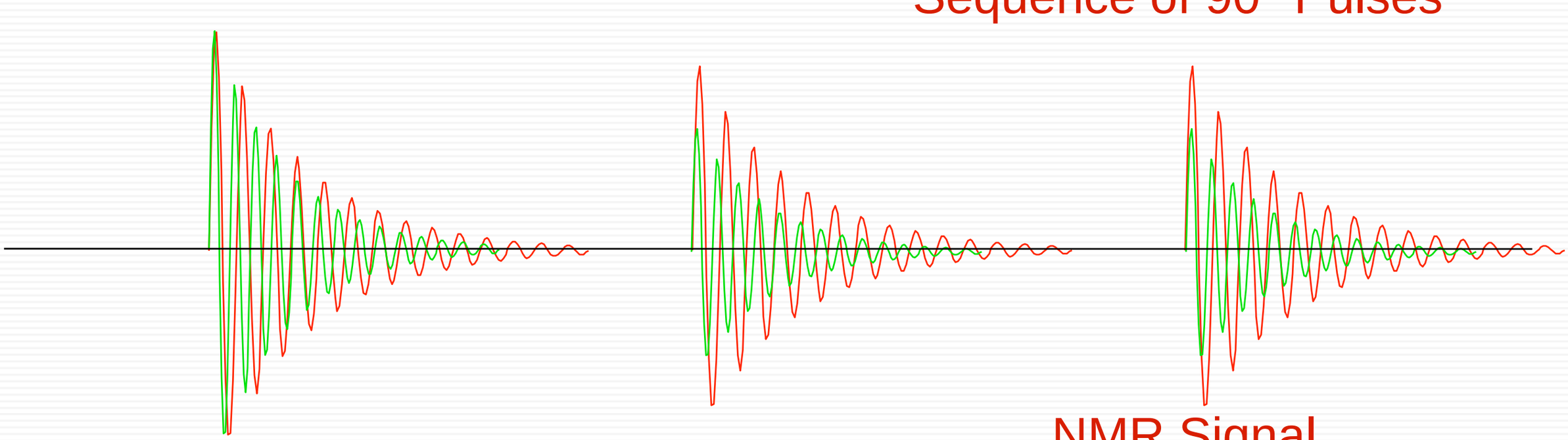
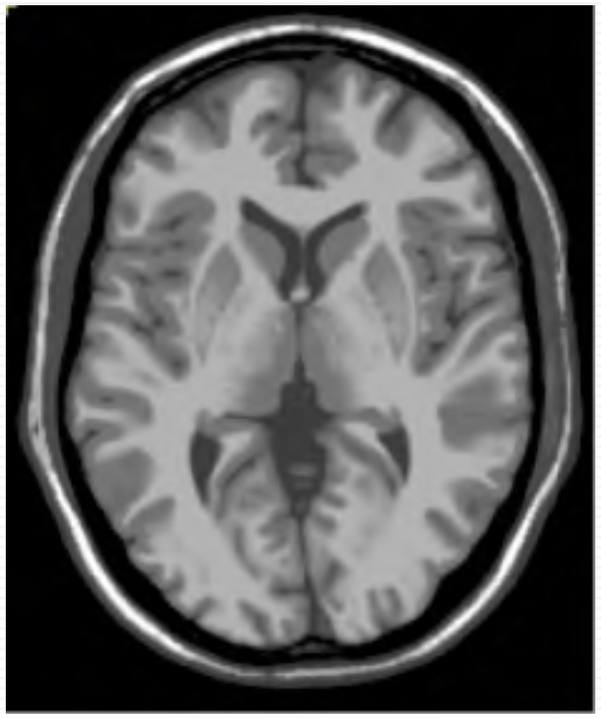


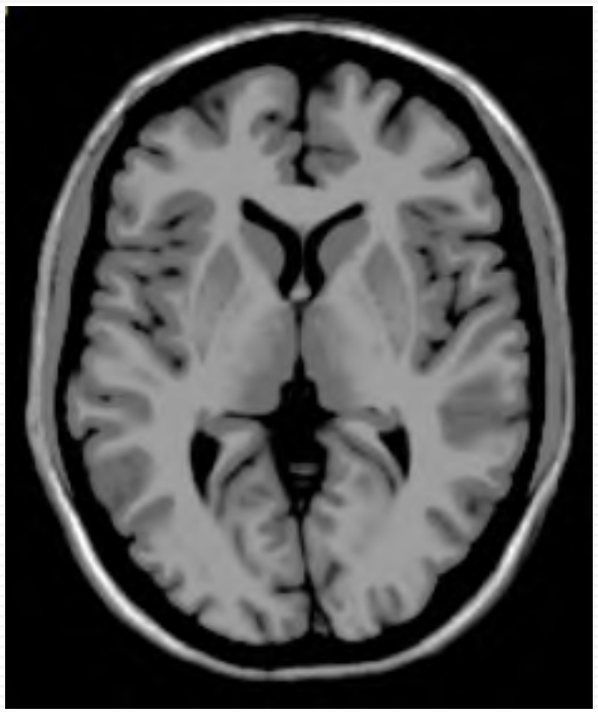
NMR Signal

tr and contrast (simulations)

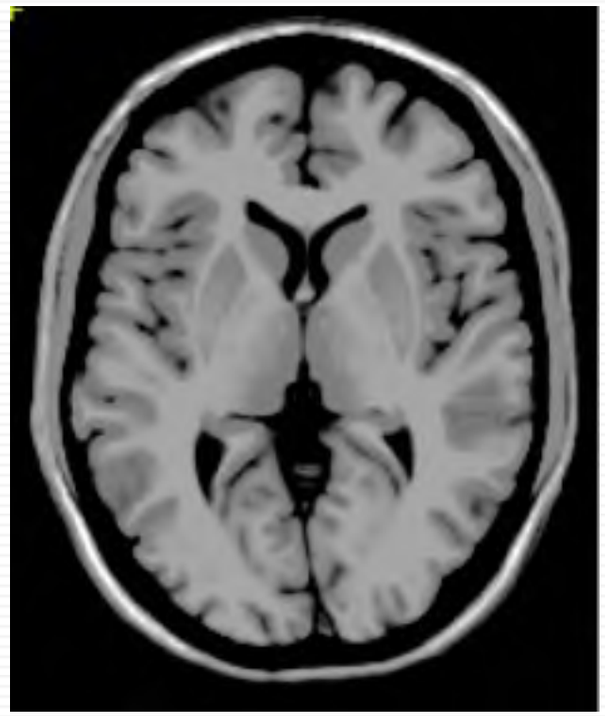
te=4



tr=100

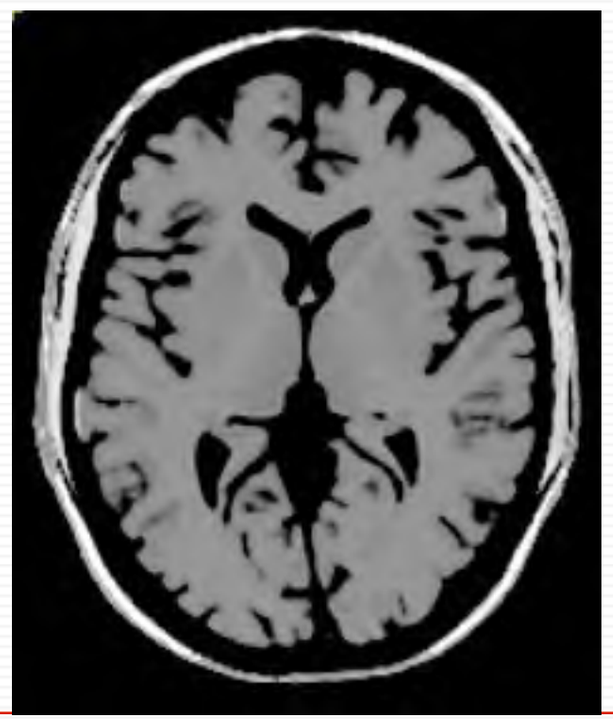


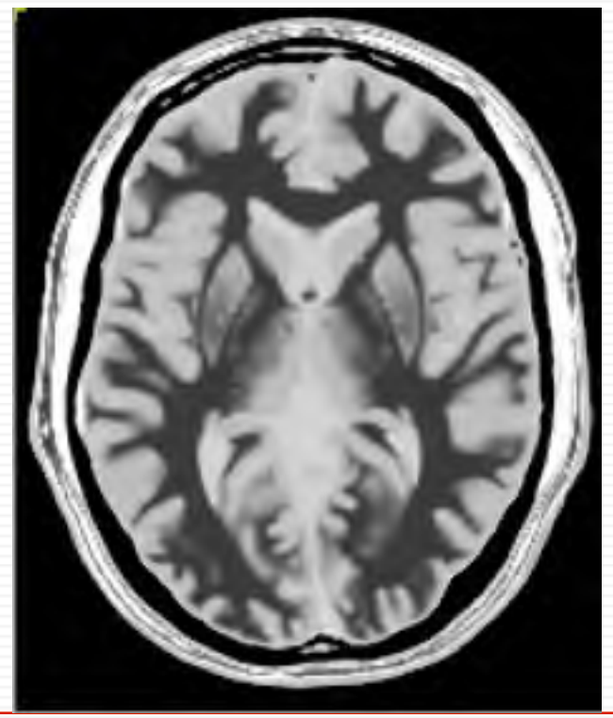
tr=400



tr=700

tr=1500



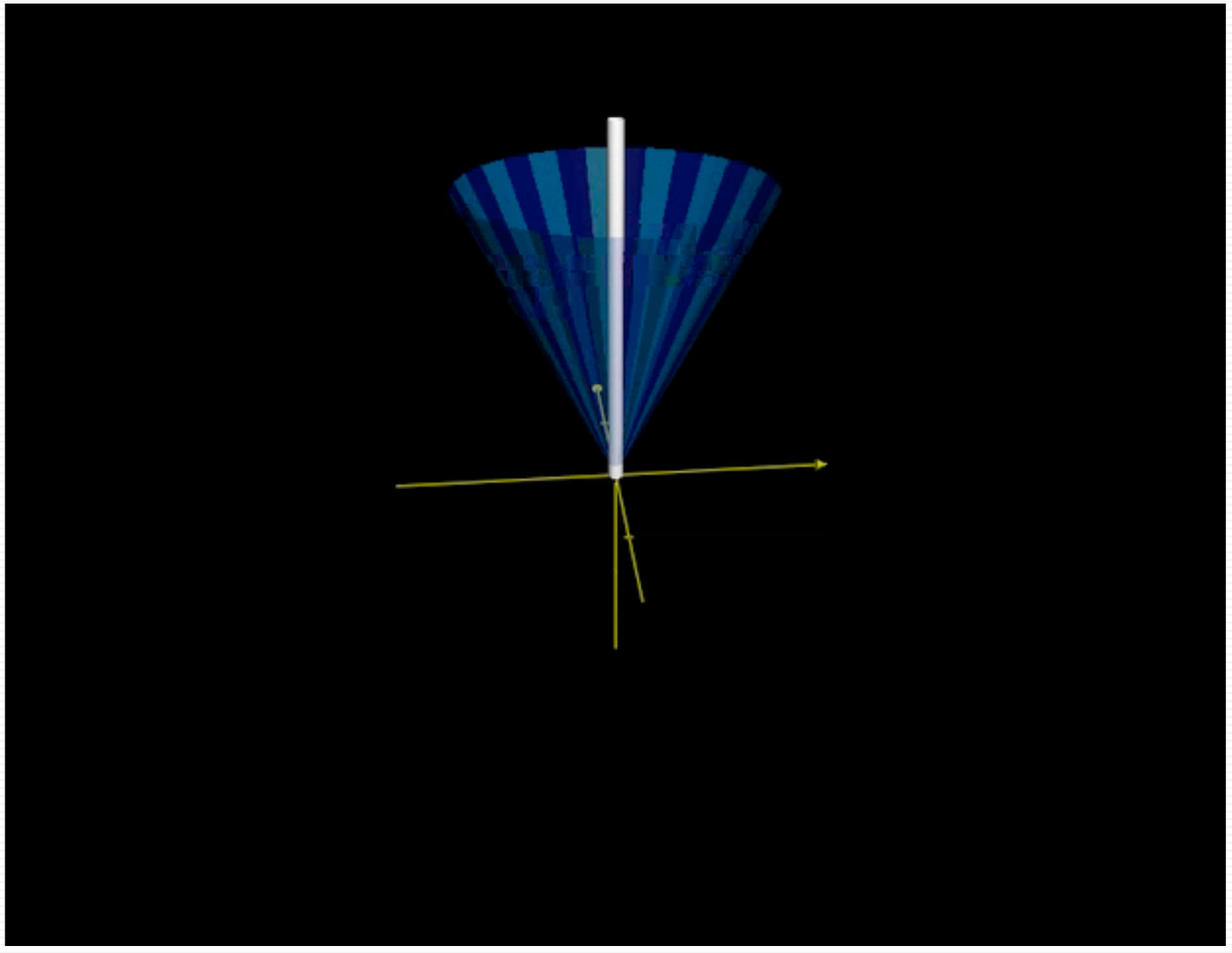


tr=5000

http://mouldy.bic.mni.mcgill.ca/cgi/bw/submit_request

Signal Decay from Dephasing

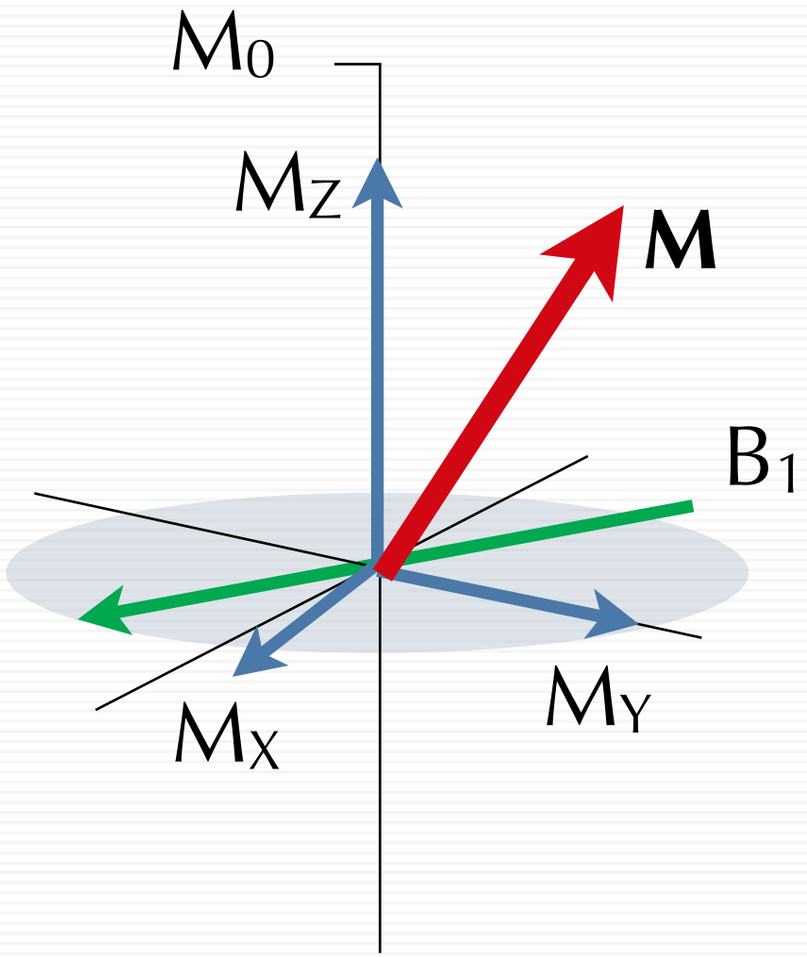
Signal Decay from Dephasing



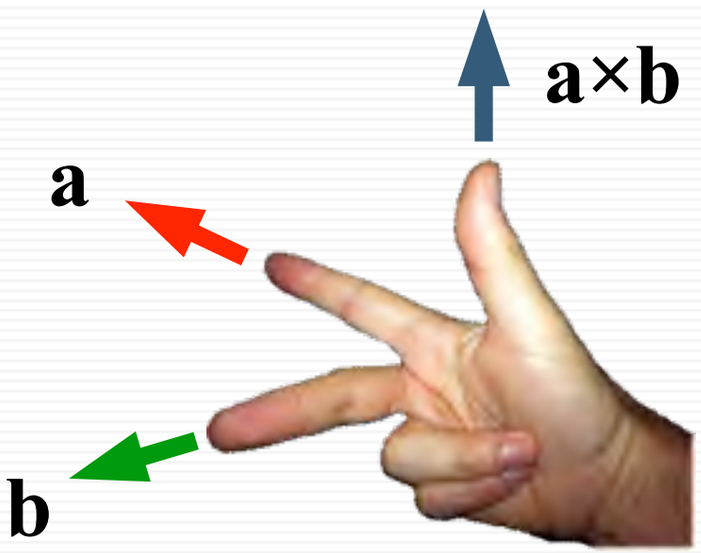
Bloch Equation

- $d\mathbf{M}/dt = \gamma\mathbf{M} \times \mathbf{B_1} + (\mathbf{M_0} - \mathbf{M_z})/T_1 - (\mathbf{M_x} + \mathbf{M_y})/T_2$

Nutation & precession — *Longitudinal relaxation* — *Transverse relaxation*



$$\mathbf{a} \times \mathbf{b} = ab\sin\theta\hat{\mathbf{n}}$$

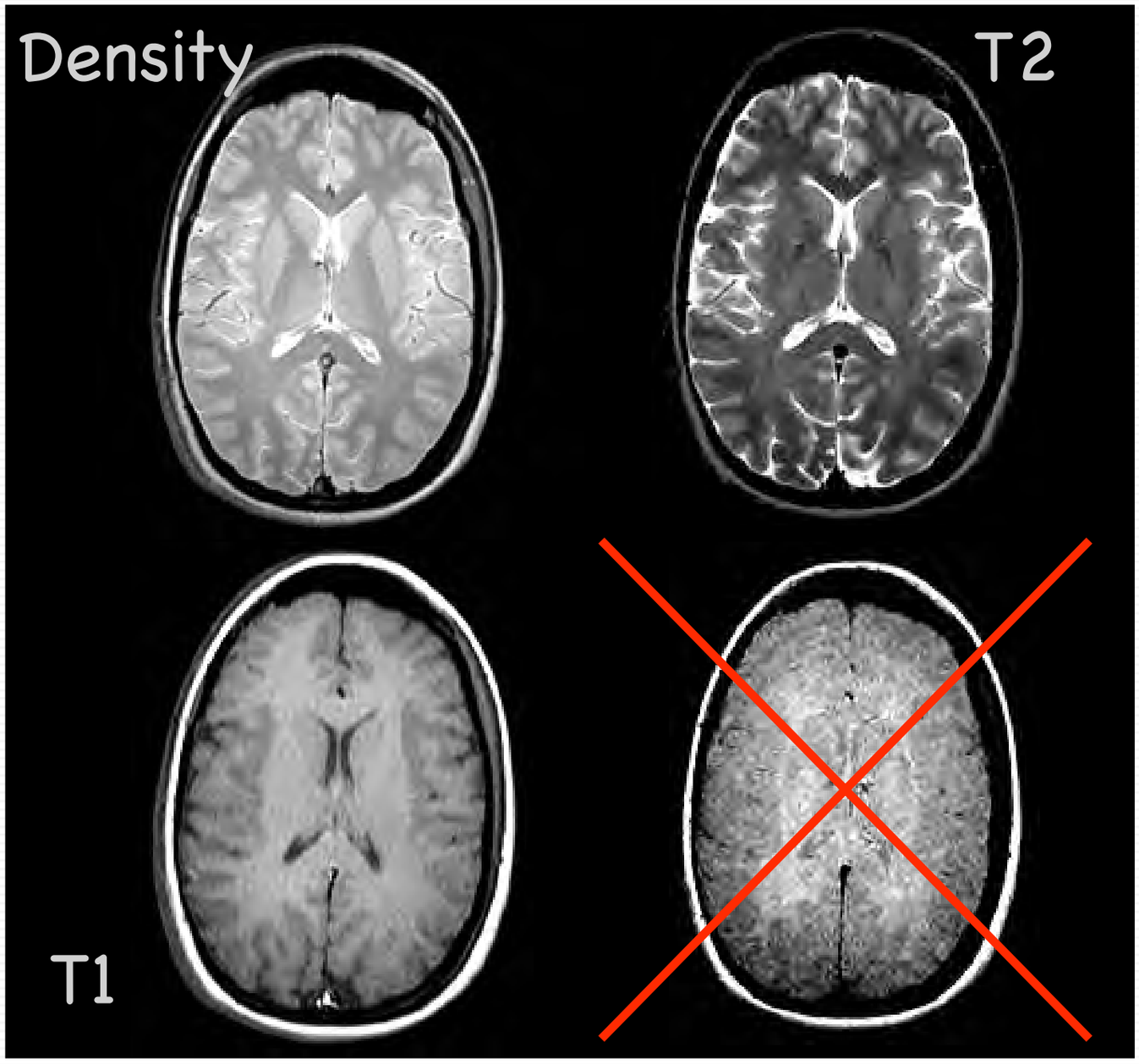


Contrast, TR and TE

$$S = k\rho M_0 (1 - e^{-tr/T1}) e^{-te/T2}$$

tr \ te	Short	Long
Long	Proton Density	T2-Weighted
Short	T1-Weighted	

Contrast, TR and TE



Long

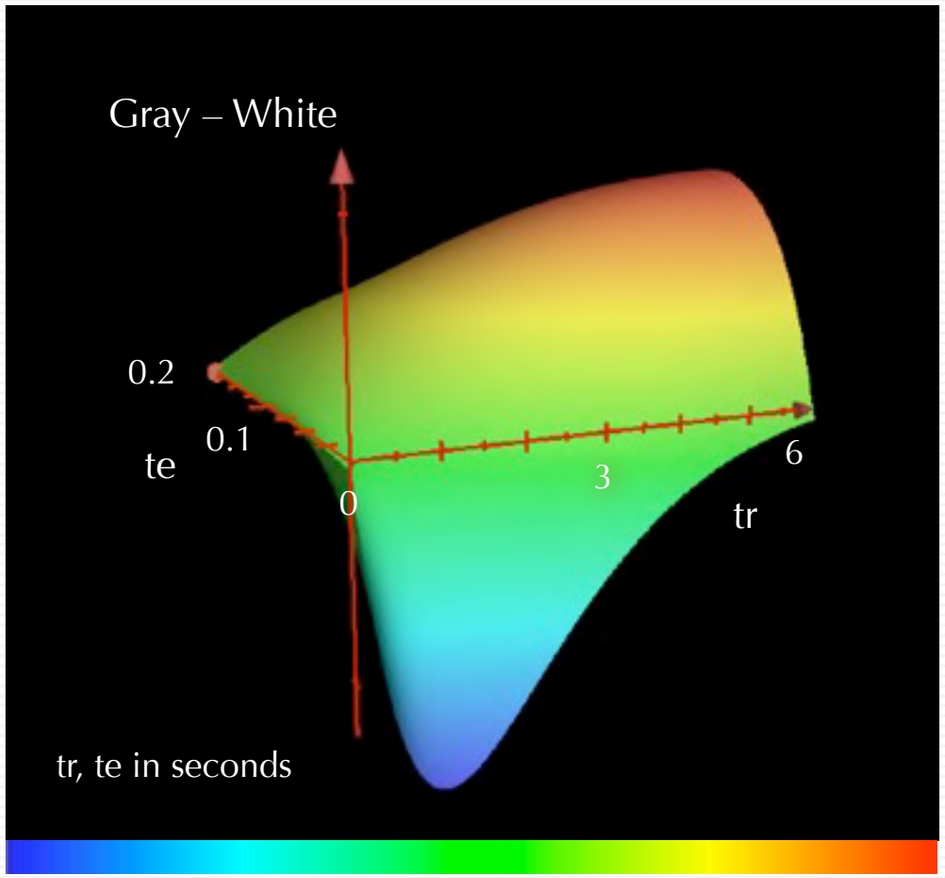
TR

Short

Short **TE** **Long**

Contrast to Noise Ratio

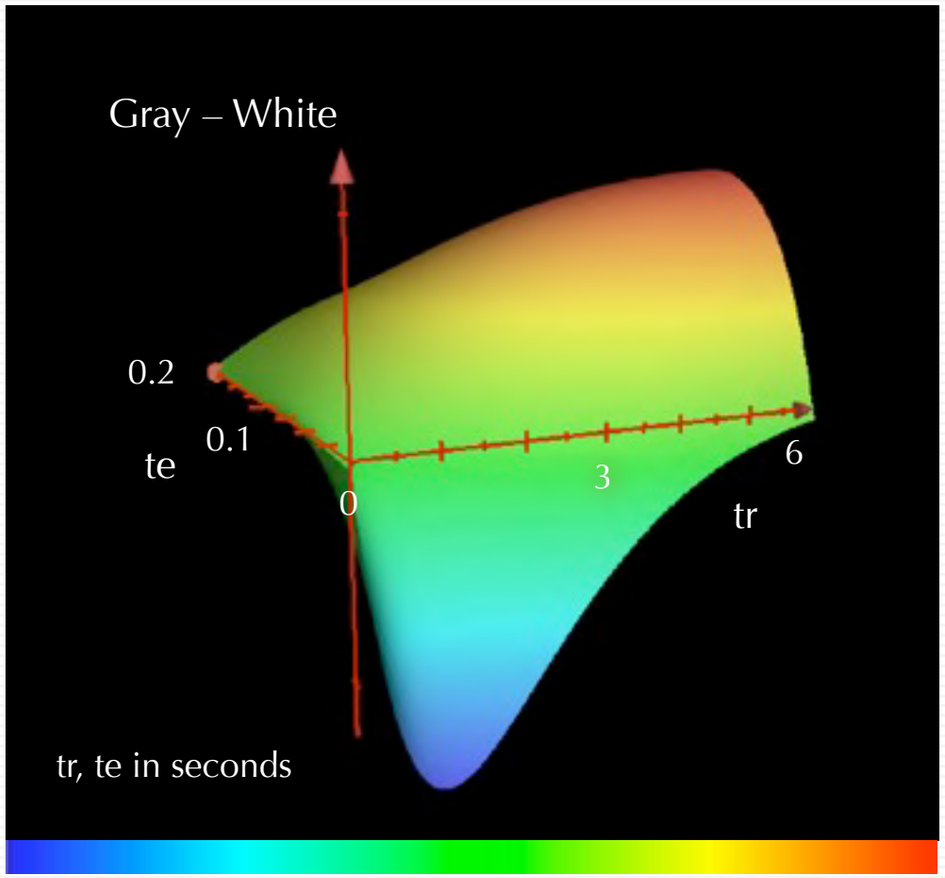
(Gray-White)

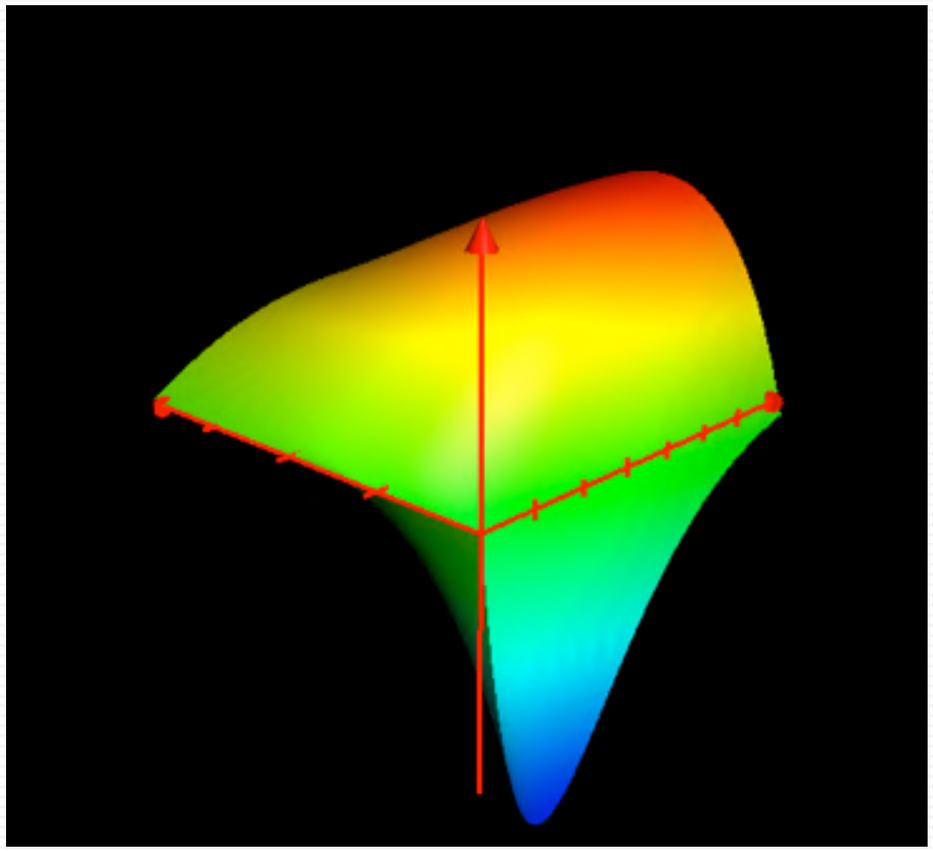


$$Contrast = [(1 - e^{-tr/1.2})e^{-te/.08}],\ gray\ matter$$
$$-[(1 - e^{-tr/1.0})e^{-te/.07}],\ white\ matter$$

Contrast to Noise Ratio

(Gray-White)





$$Contrast = [(1 - e^{-tr/1.2})e^{-te/.08}],\ gray\ matter$$

$$-[(1 - e^{-tr/1.0})e^{-te/.07}],\ white\ matter$$

T2*

the Observed Transverse Relaxation Rate, T2*, is the sum of several components:

$$\frac{1}{T2^*} = \frac{1}{T2} + \frac{1}{T2'} + \frac{1}{T2_D} + \ldots$$

Molecular

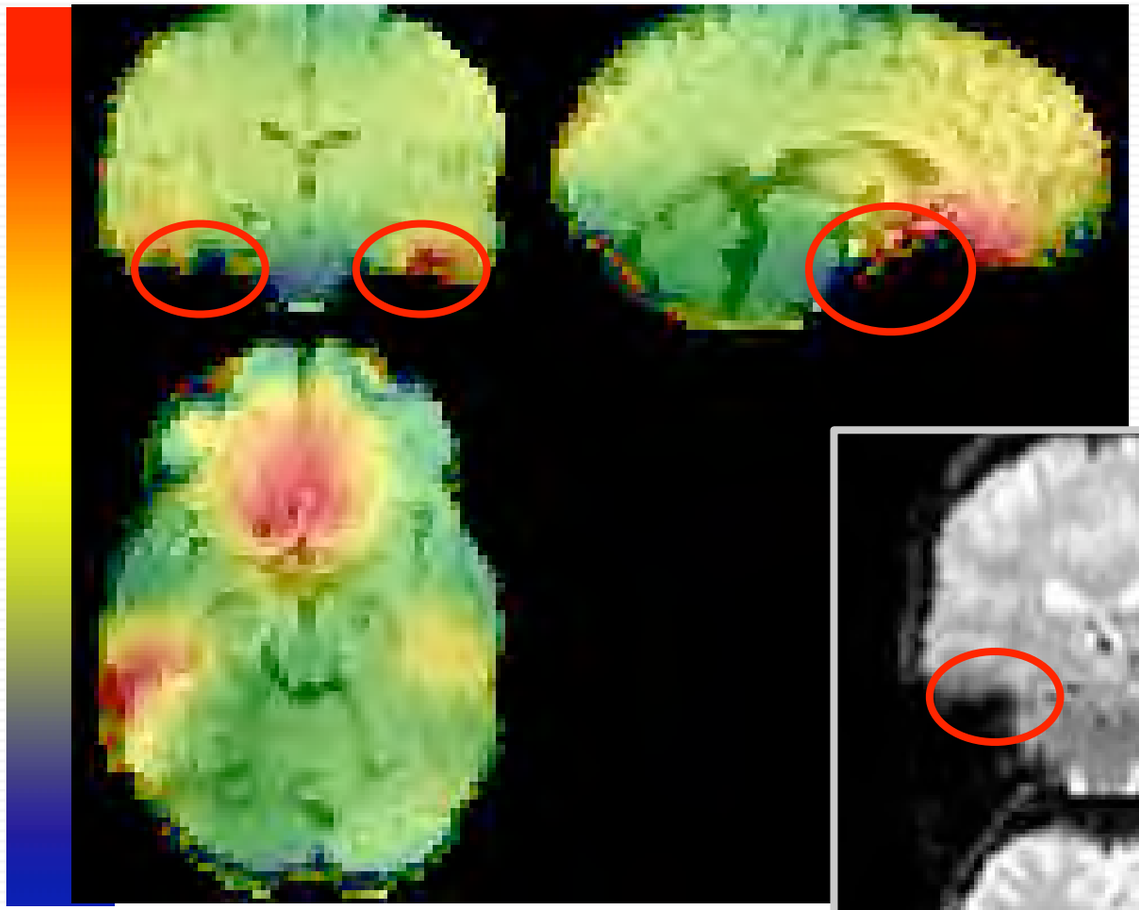
Field Inhomogeneity

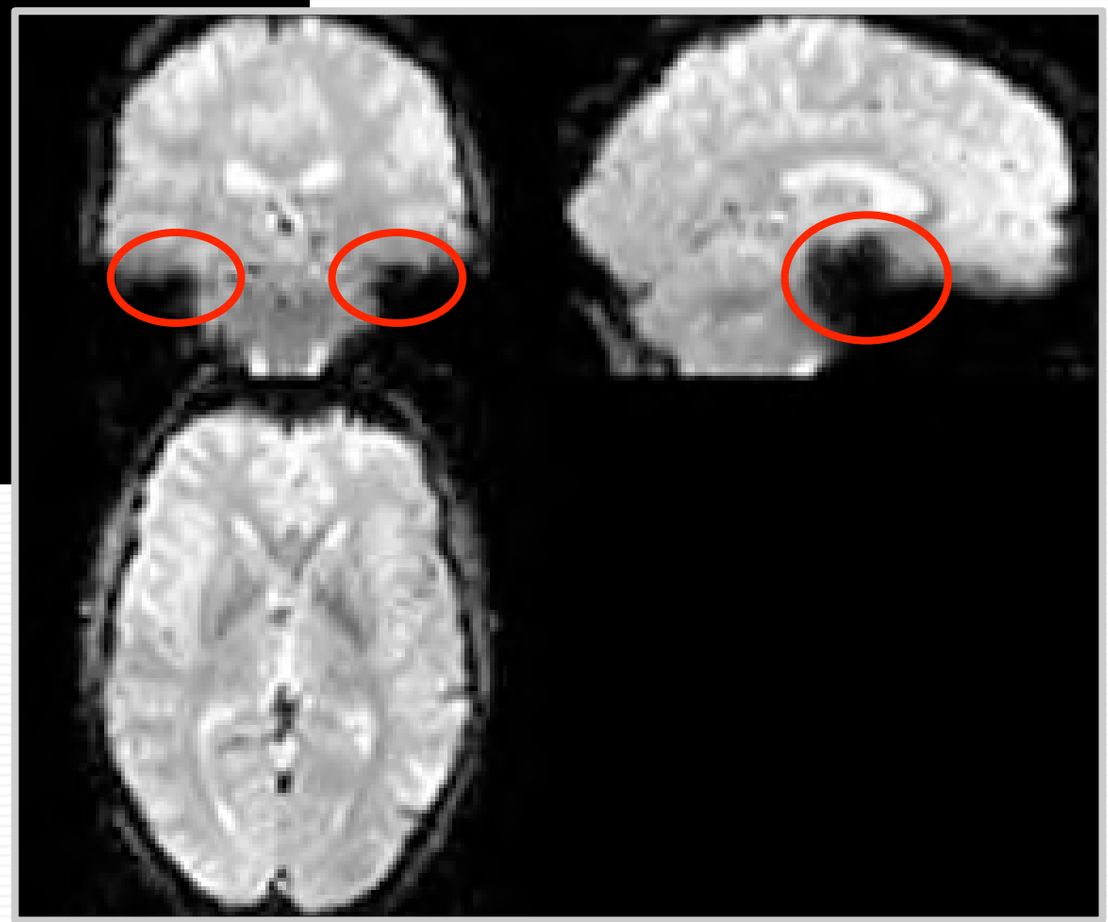
Diffusion

Local field Variations Result in Signal Loss

+3 ppm

-3 ppm

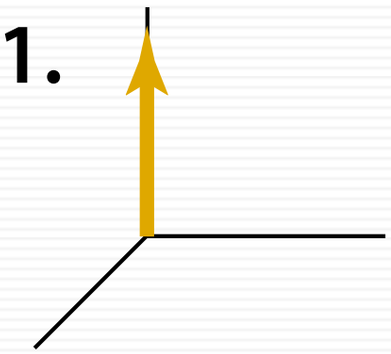




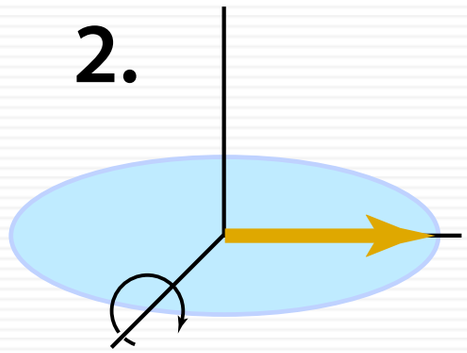
Hahn Spin Echo

Erwin Hahn

1.

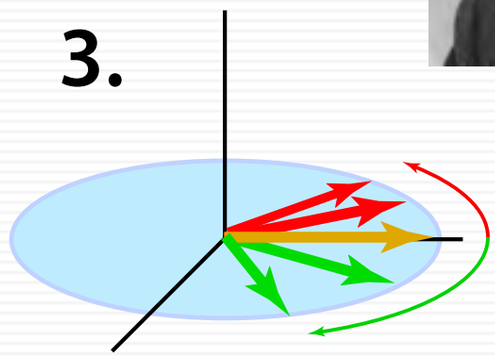


2.



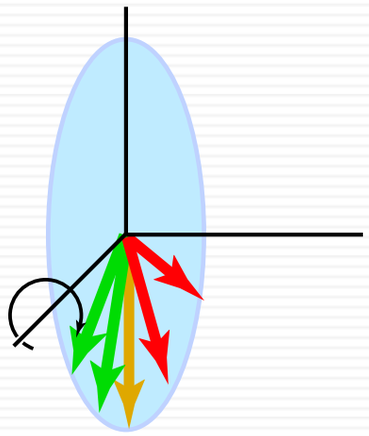
90° pulse

3.



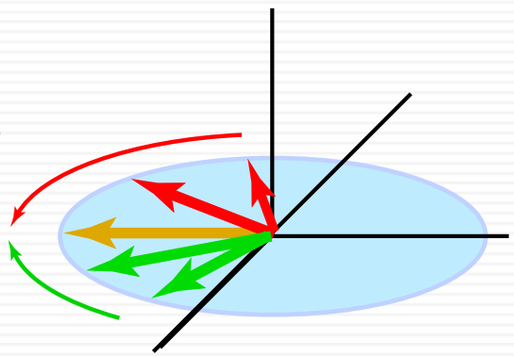
dephasing

4.



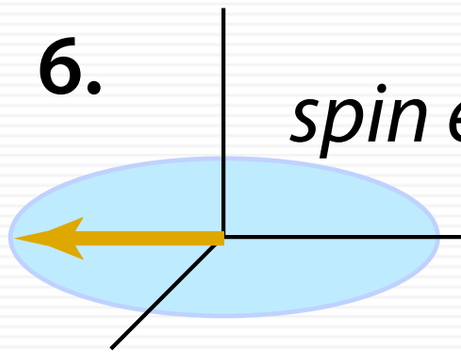
180° pulse

5.



rephasing

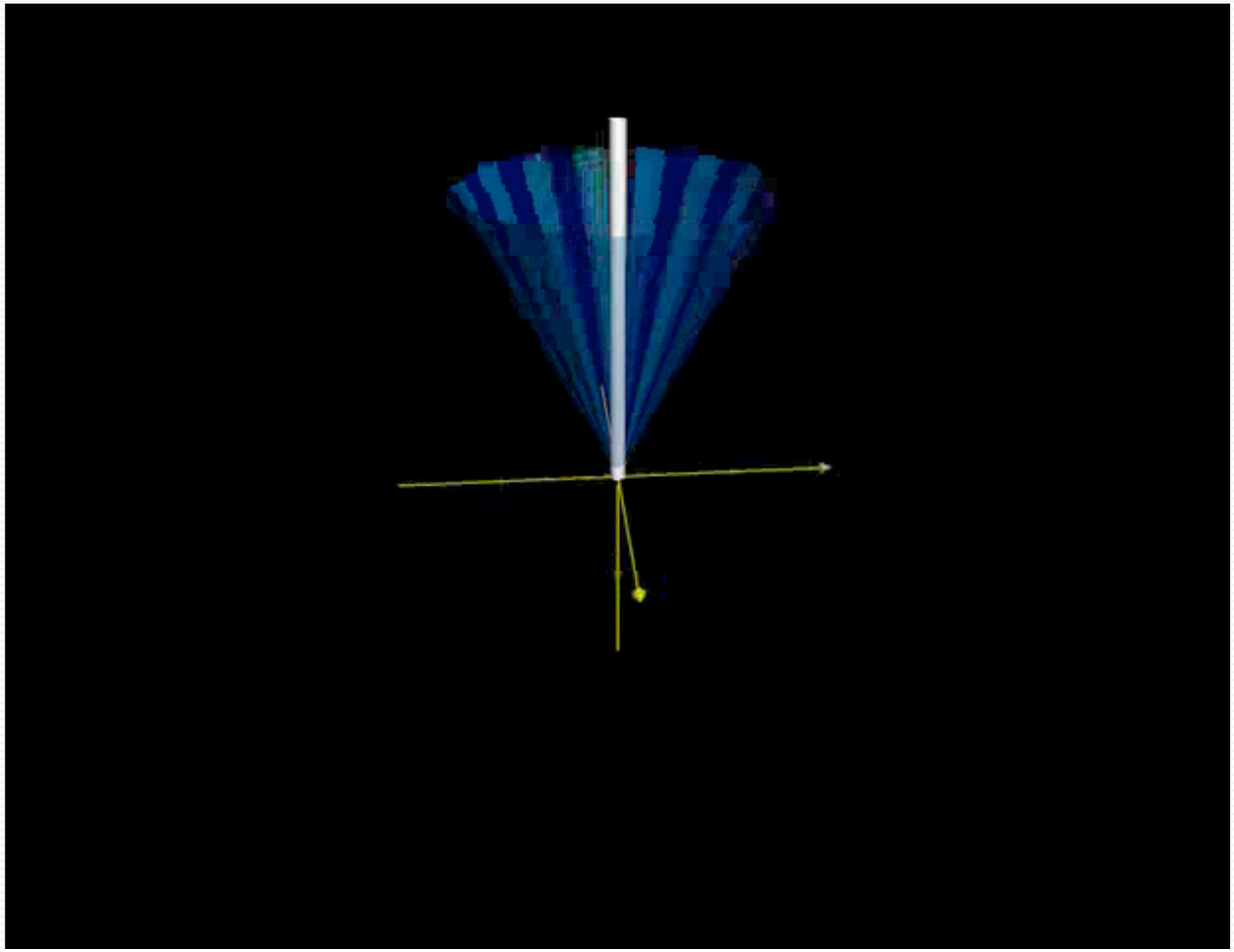
6.



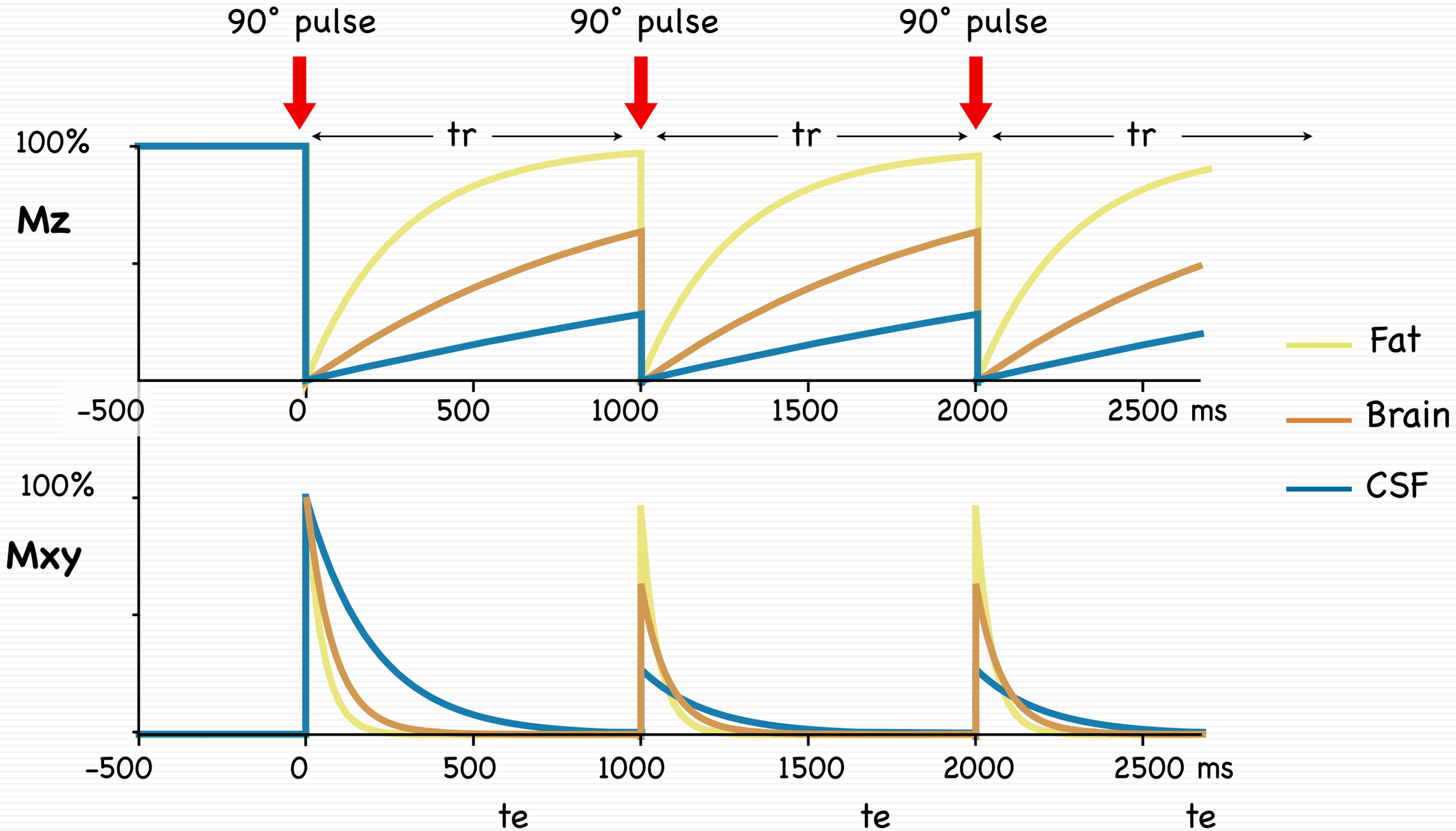
spin echo

Hahn Spin Echo

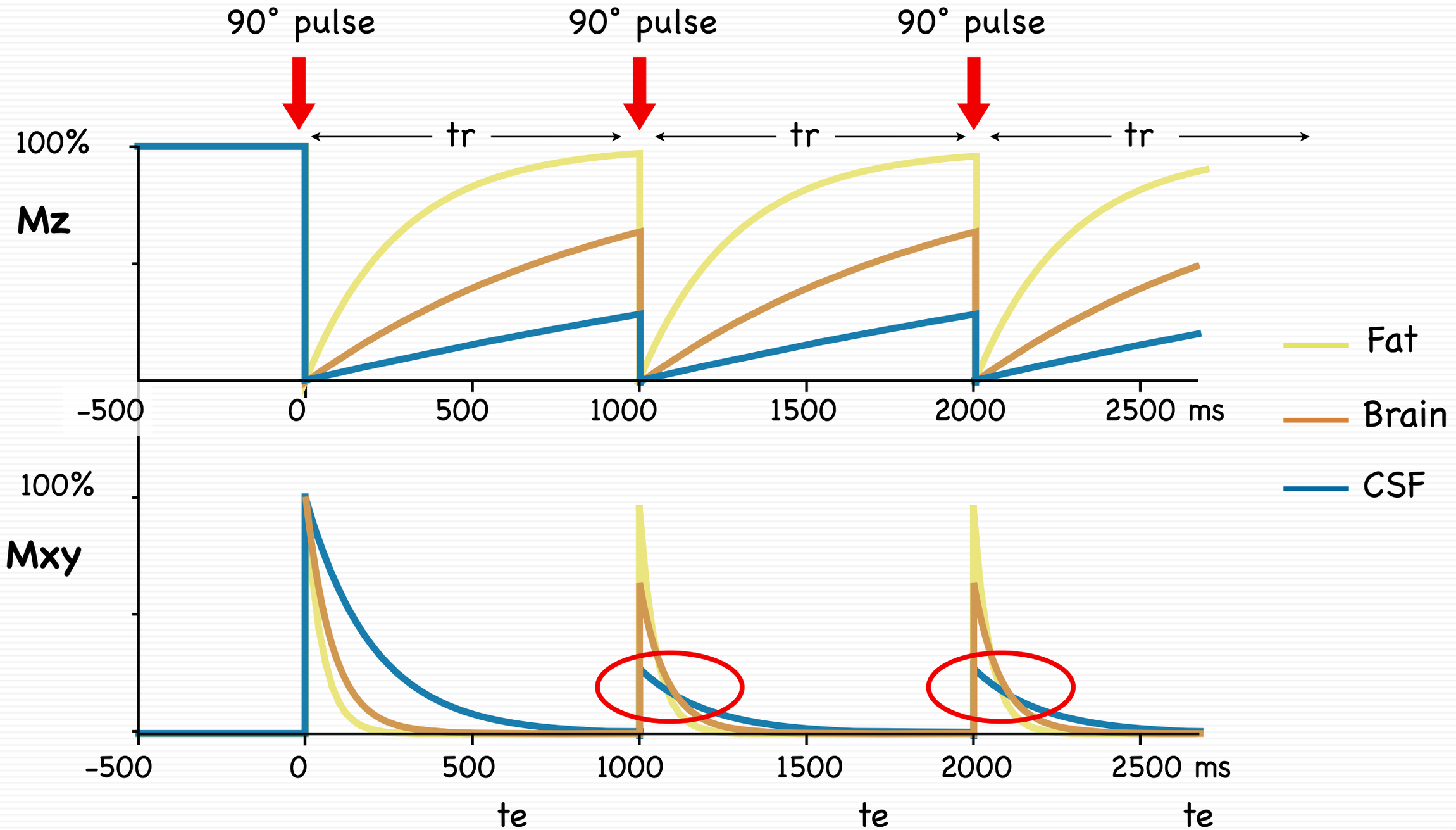
Hahn Spin Echo



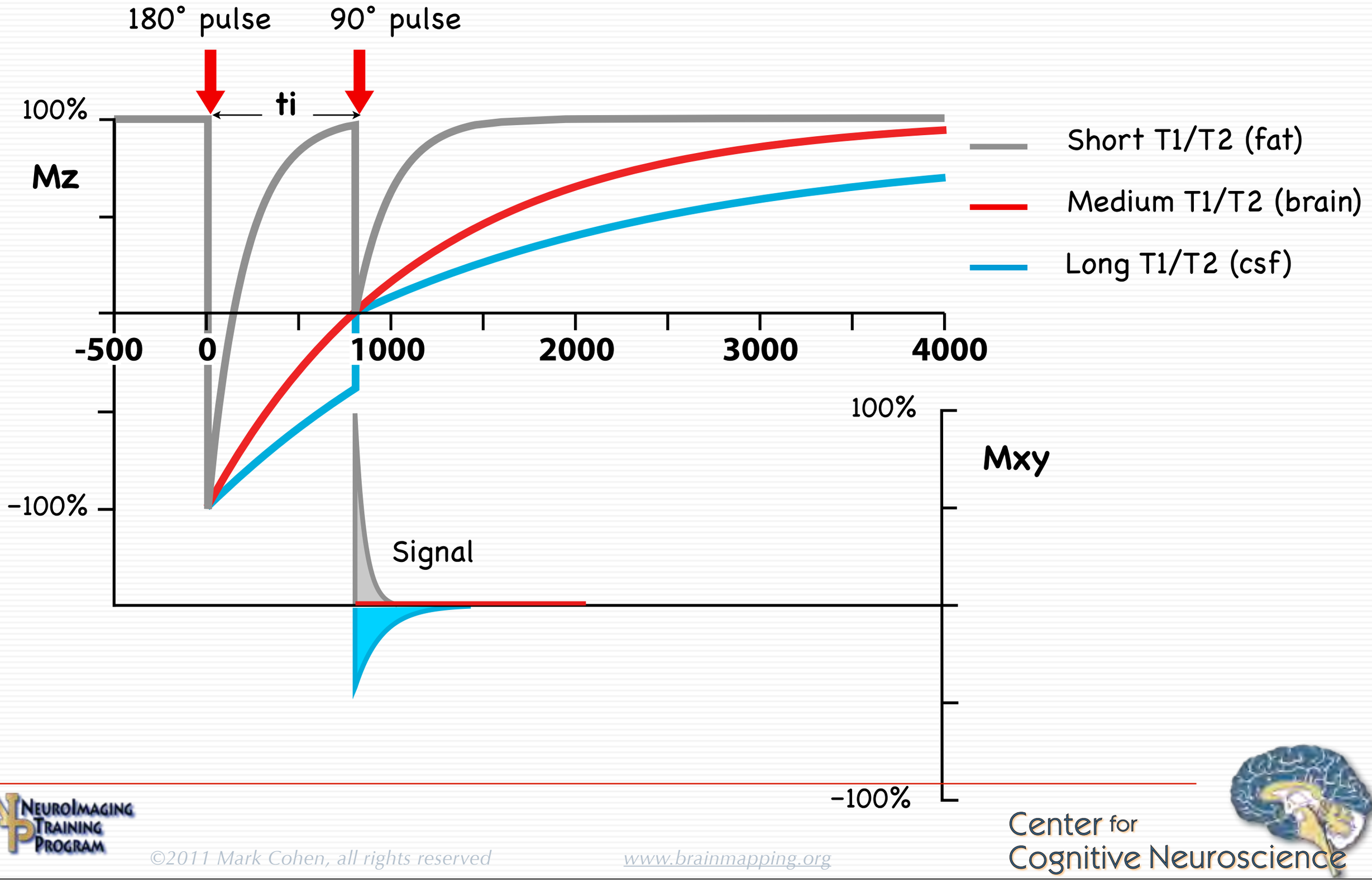
Longitudinal & Transverse Magnetization



Longitudinal & Transverse Magnetization



Inversion Recovery

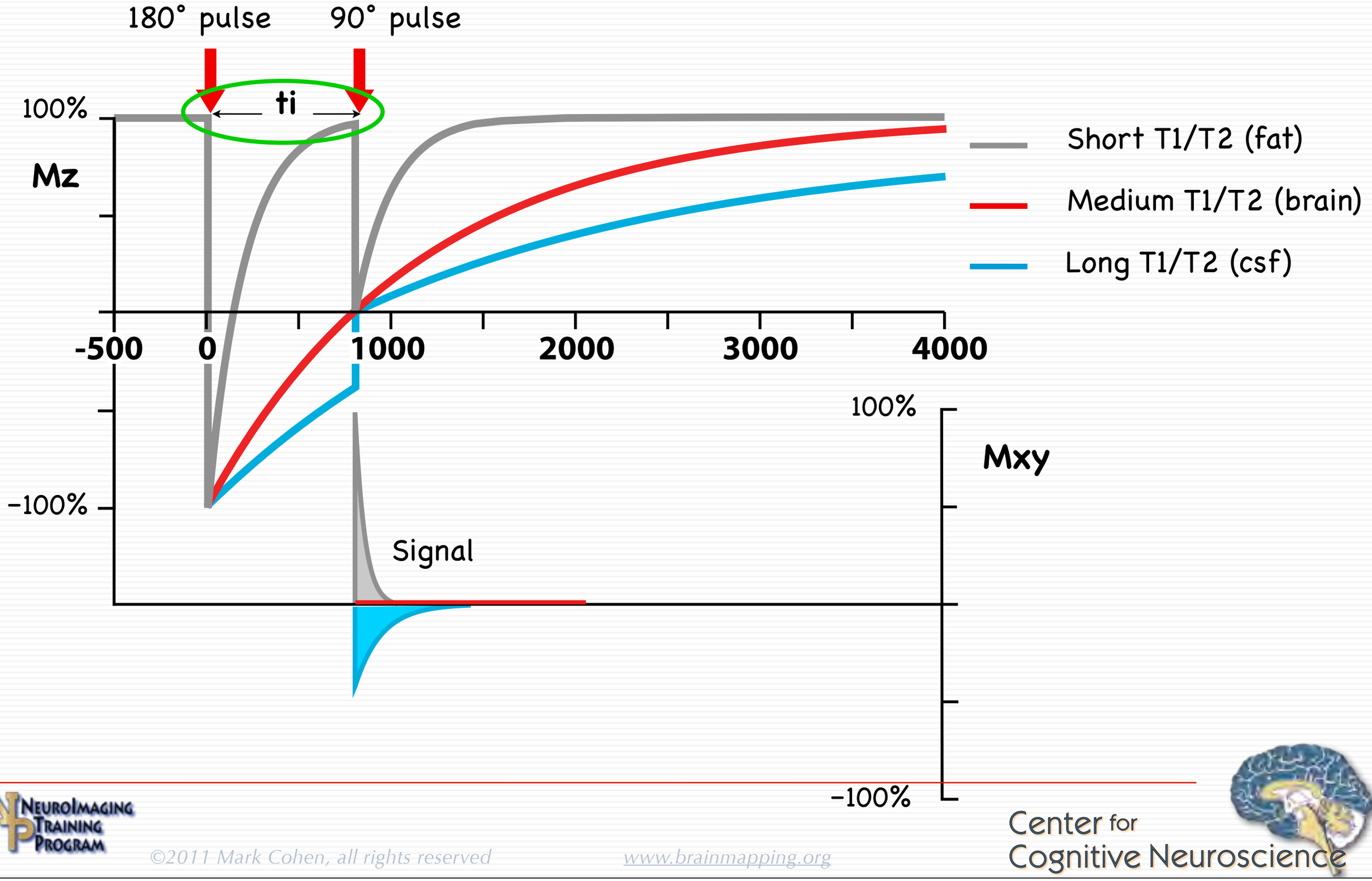


NeuroImaging Training Program

www.brainmapping.org

Center for Cognitive Neuroscience

Inversion Recovery



NeuroImaging Training Program

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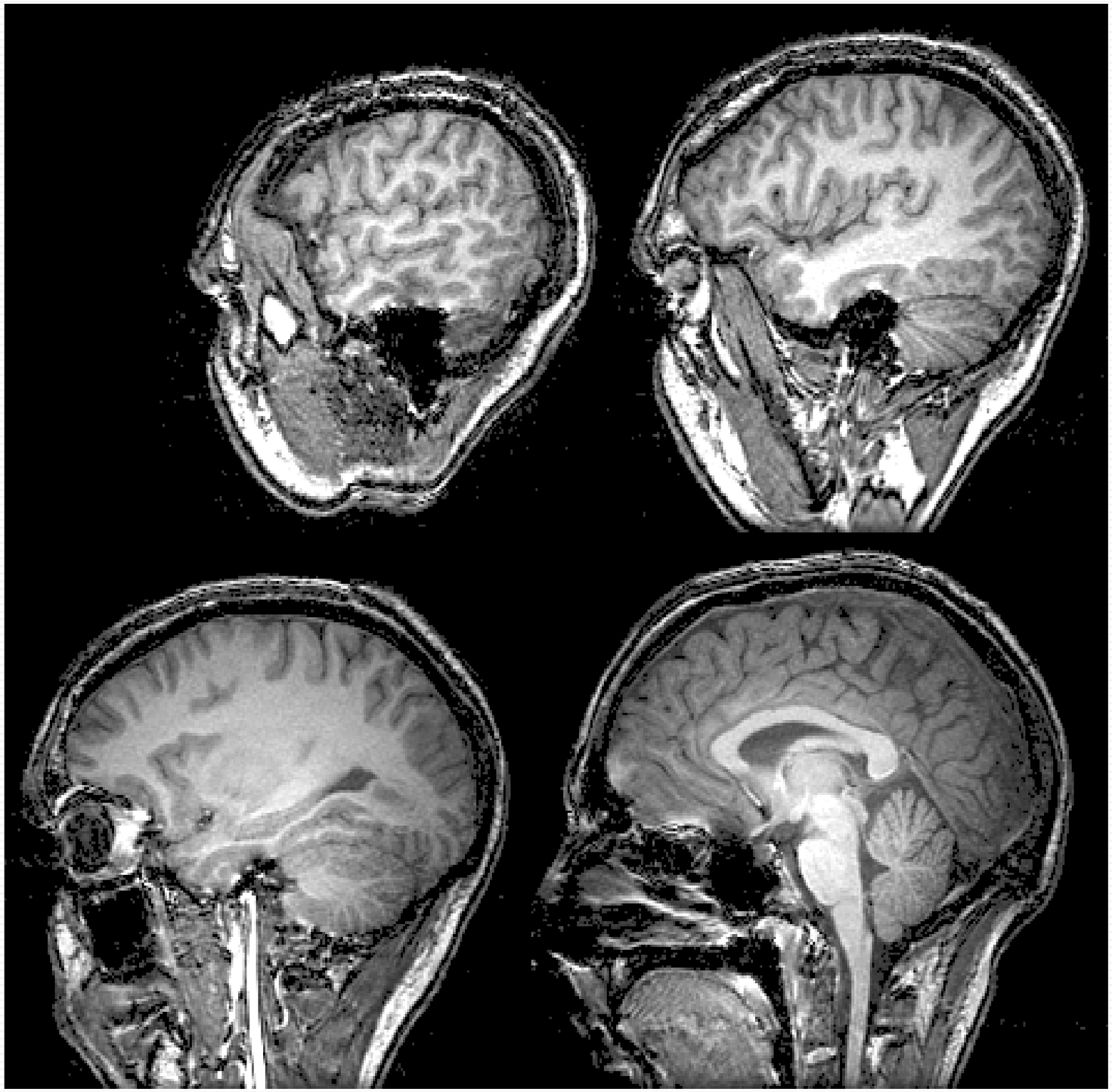
Center for Cognitive Neuroscience

3D T1 Images

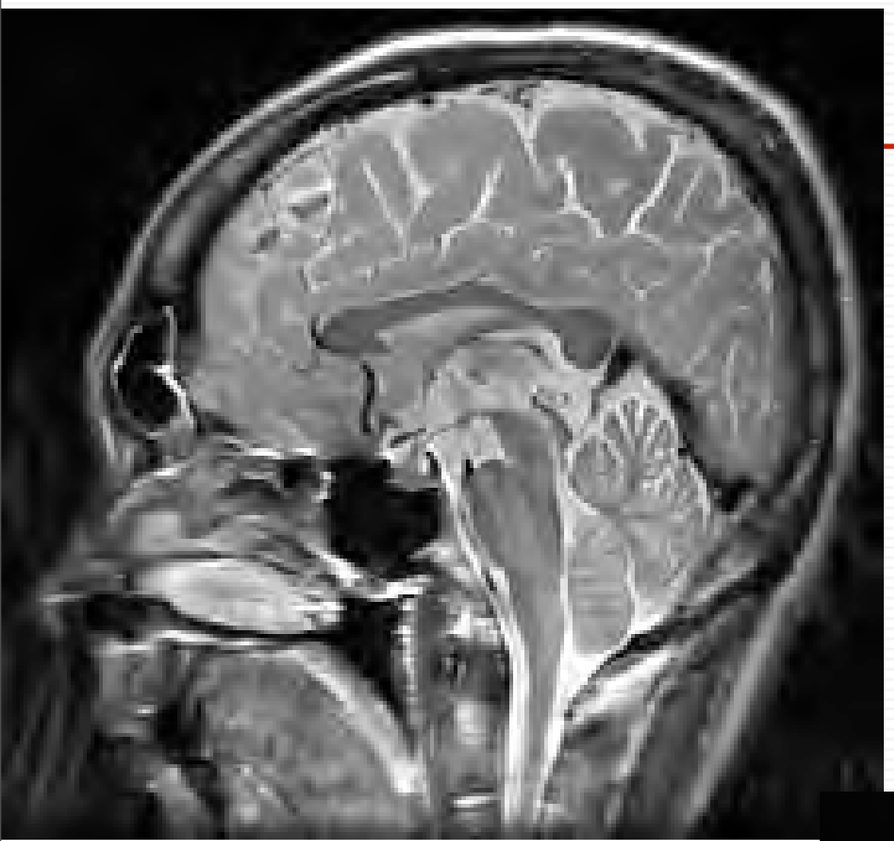
TE = 3.2
TR = 14.4
124 slices
1.25 mm thick
1 NEX
Flip Angle 20°
TI = 500



Sample Data Set (normal)

Fast Spin Echo

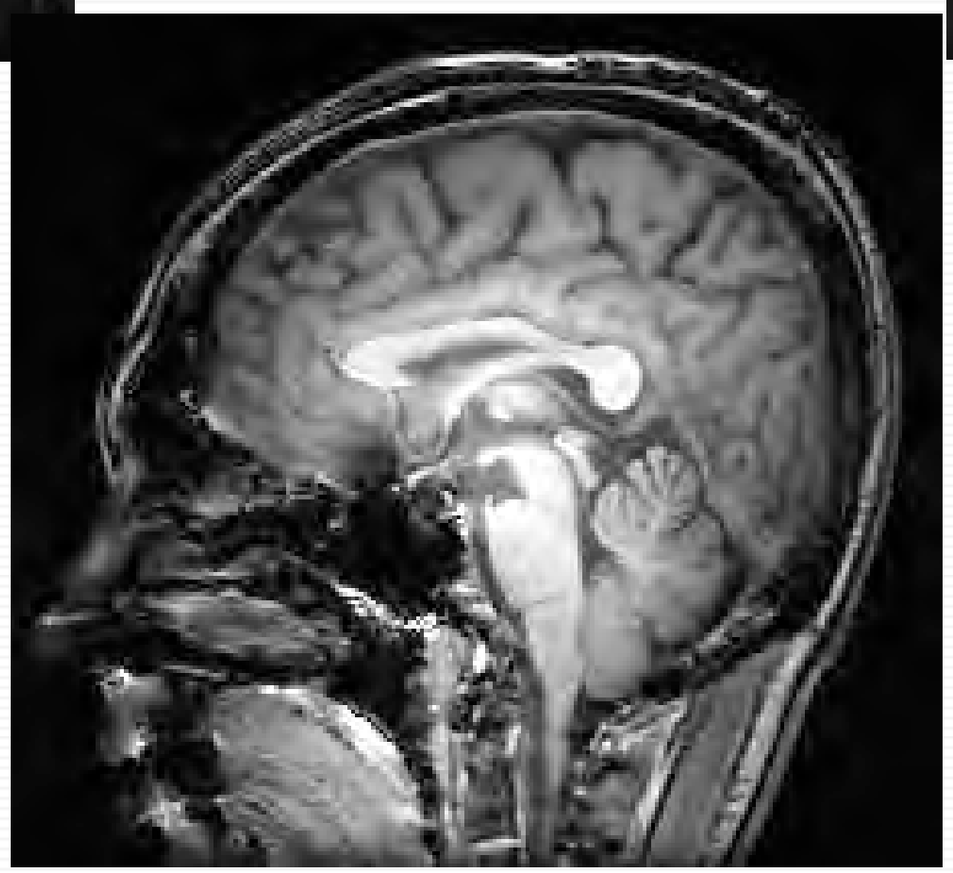
3 mm Slices



Proton Density

3D IR-SPGR

TE = 3.2, TI = 700



T2-weighted

T1-weighted

Sample Data Set (normal)

MR Formulæ

Felix Bloch

Spin Echo Signal = $k\rho M_0(1 - e^{-tr/T1})e^{-te/T2}$

Inversion Recovery Signal =

$$k\rho M_0(1 - 2e^{-ti/T1} + e^{-tr/T1})e^{-te/T2}$$

ρ is the proton density

k represents instrument effects

The "Bloch" Equation:

$$d\mathbf{M}/dt = \gamma\mathbf{M} \times \mathbf{B_1} + (\mathbf{M}_0 - \mathbf{M}_z)/T_1 - (\mathbf{M}_x + \mathbf{M}_y)/T_2$$

MRI Contrast Summary

…next: Making an Image

MRI Contrast Summary

- Pulses of Rotating Magnetic Fields (RF) Convert Nuclear Magnetization to Signal

…next: Making an Image

MRI Contrast Summary

- Pulses of Rotating Magnetic Fields (RF) Convert Nuclear Magnetization to Signal
- RF Pulses Add Energy by Displacing Longitudinal Equilibrium

…next: Making an Image

MRI Contrast Summary

- Pulses of Rotating Magnetic Fields (RF) Convert Nuclear Magnetization to Signal
- RF Pulses Add Energy by Displacing Longitudinal Equilibrium
- Contribution of Intrinsic Tissue Properties T1 and T2 Manipulated by Experimenter controlled timing: tr and te respectively.

…next: Making an Image

MRI Contrast Summary

- Pulses of Rotating Magnetic Fields (RF) Convert Nuclear Magnetization to Signal
- RF Pulses Add Energy by Displacing Longitudinal Equilibrium
- Contribution of Intrinsic Tissue Properties T1 and T2 Manipulated by Experimenter controlled timing: tr and te respectively.
- Typically, $0.05 < T1 < 3s$ and
  $0.005 < T2 < .3s$ for body tissues.

…next: Making an Image

The Plan

- The Magnetic Resonance Phenomenon & Contrast (30)
- Spatial Encoding (26)
- The "Pulse Sequence" Rules Everything (3)

Seventh Inning Stretch

- Fast Imaging (14)
- Functional MRI (18)
- Diffusion and Summary (9)

- Image Quality and Artifacts (48)

MR Spatial Encoding

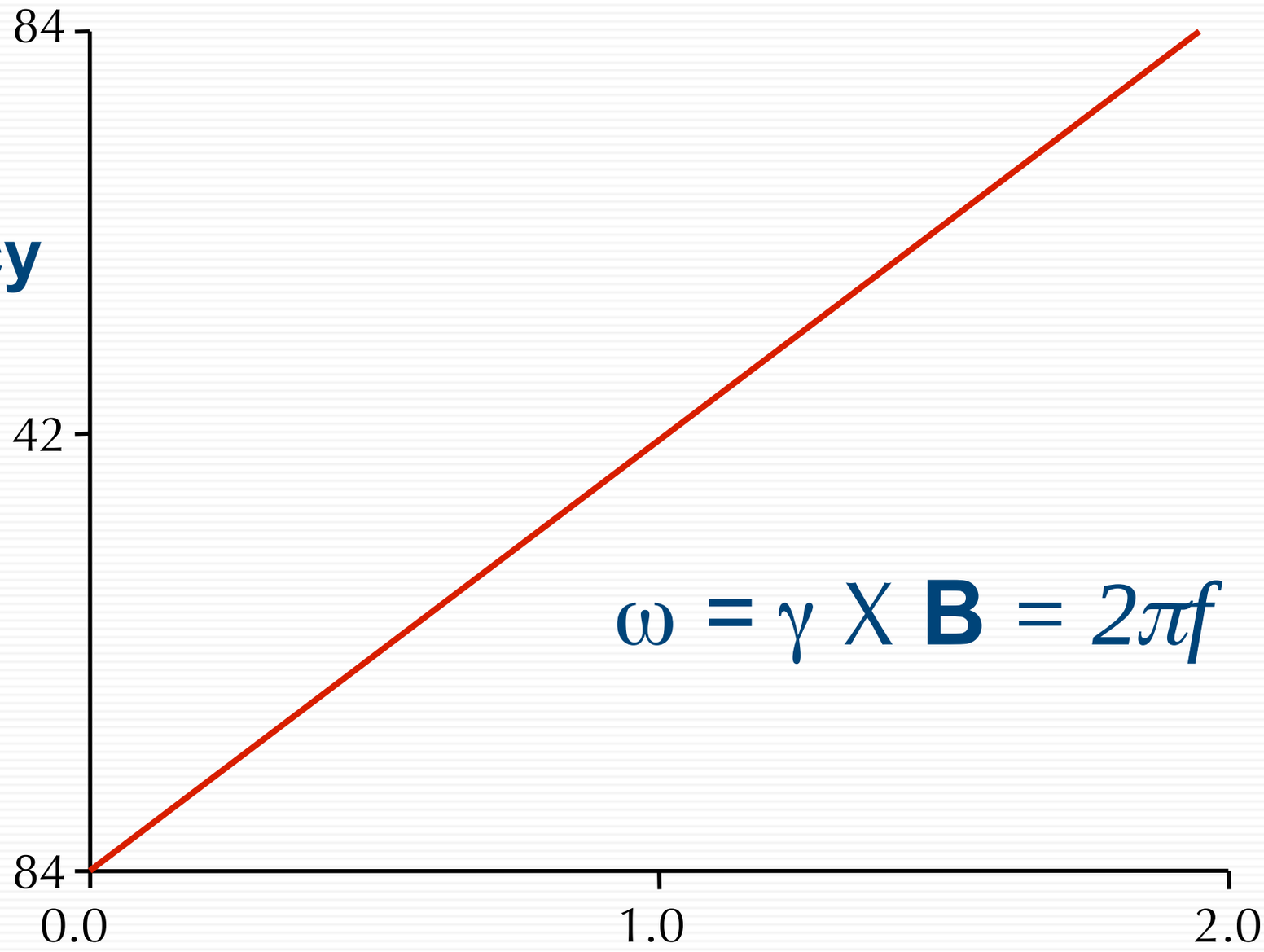
MR Spatial Encoding

- Most MR Spatial encoding is based on a single concept:
  - If the **Magnetic Field** varies by location, the **MR Frequency** will vary by location,
  - Therefore: We can *find the location by measuring the frequency*.

MR Spatial Encoding

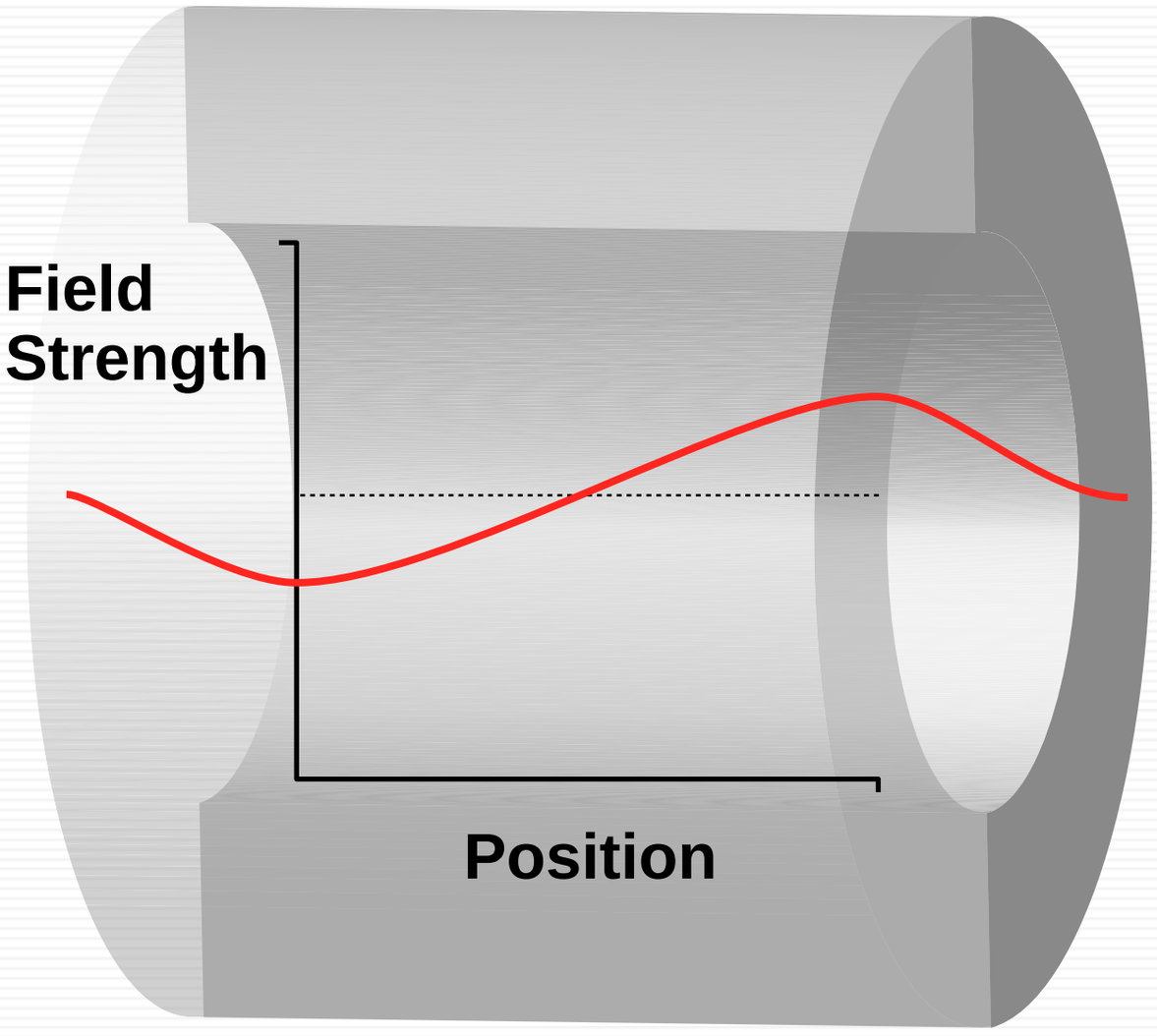
- Most MR Spatial encoding is based on a single concept:
  - If the **Magnetic Field** varies by location, the **MR Frequency** will vary by location,
  - Therefore: We can *find the location by measuring the frequency*.
- Newer methods (e.g., GRAPPA, SENSE) also find location by signal strength as a function of antenna distance

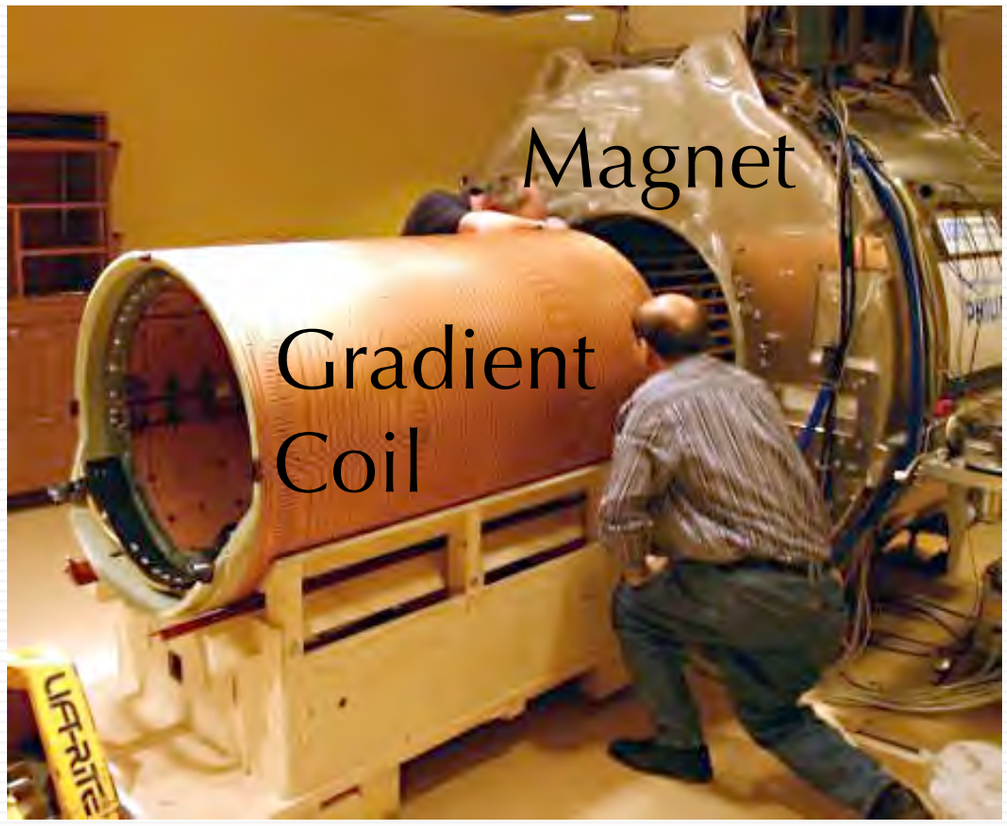
The Larmor Relation



$$\omega = \gamma \times \mathbf{B} = 2\pi f$$

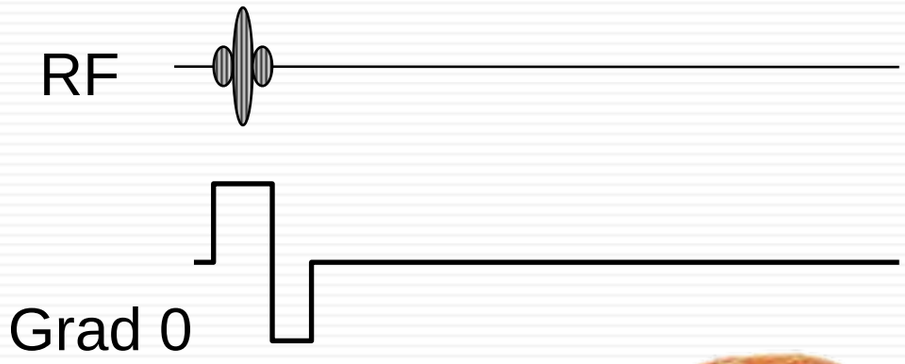
Magnetic Field Gradients





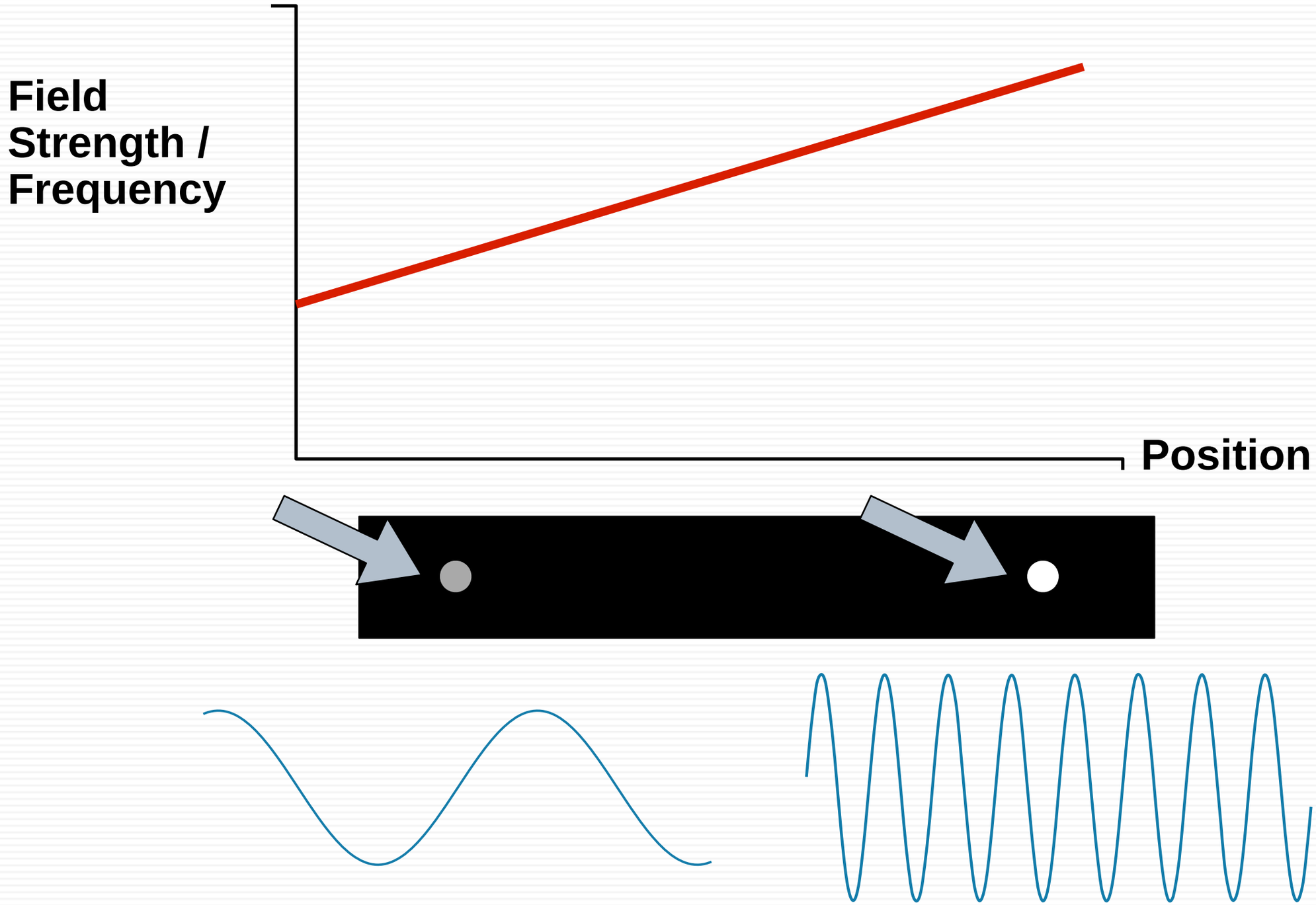
MRI Instrument in Cross Section

Frequency Selective Excitation



Sir Peter Mansfield

A 1D Image



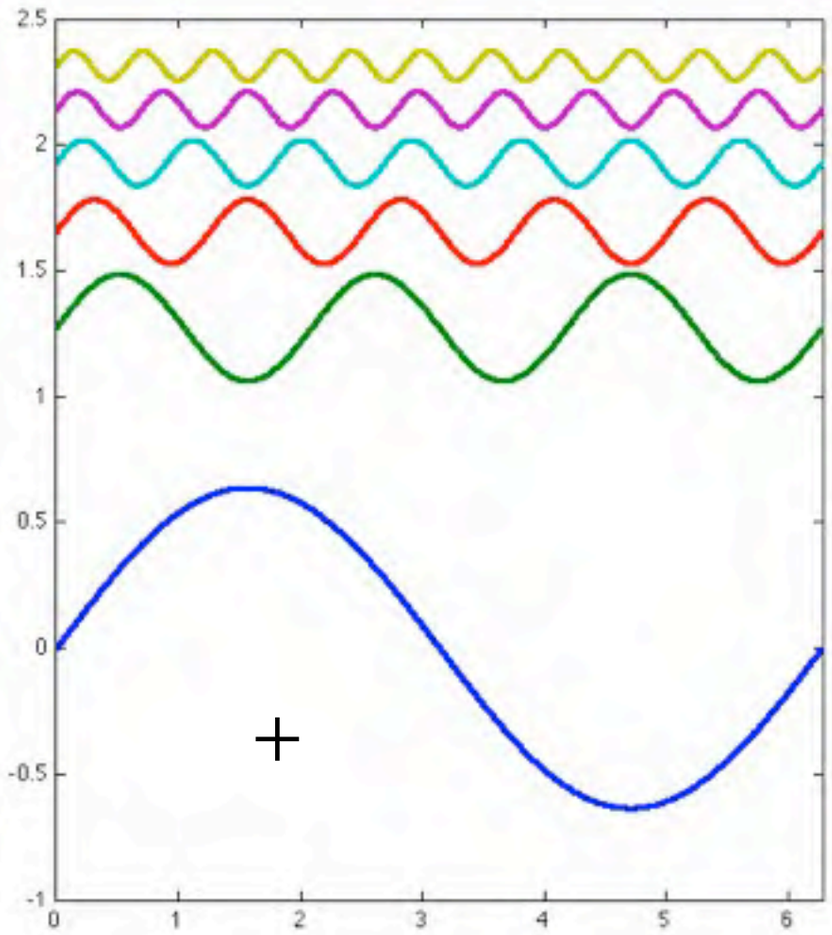
Fourier Sum example

$\sin(11t)$

$\vdots$

$\sin(5t)$

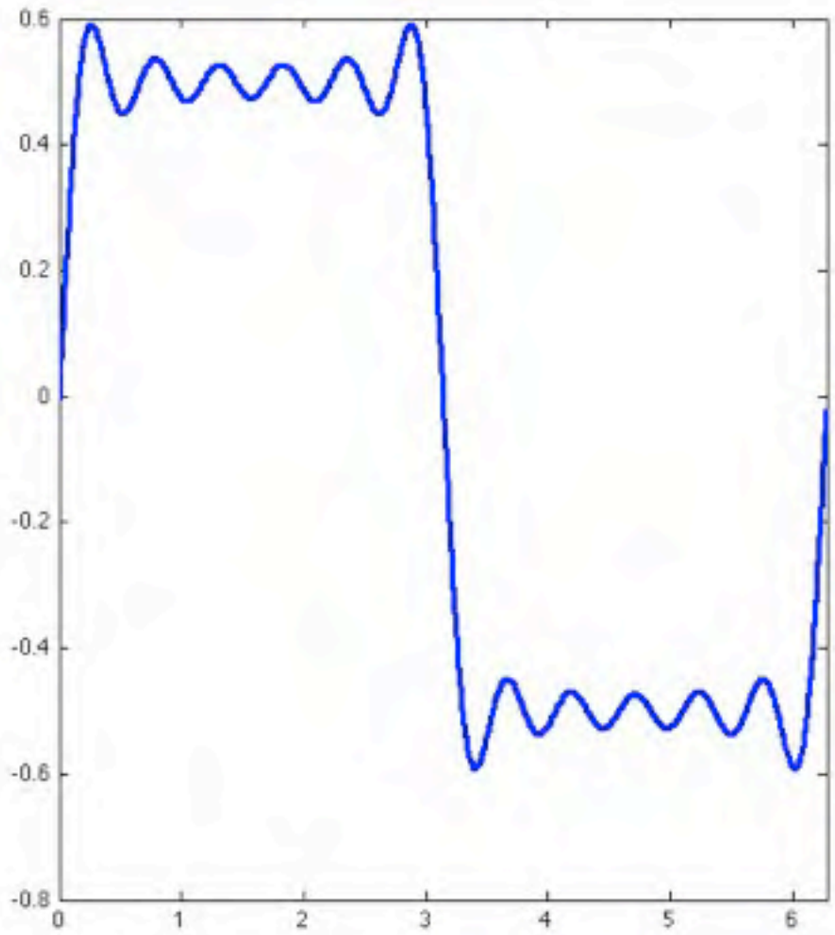
$\sin(3t)$



$\sin(t)$

+

=



Fourier Sum example

$\sin(11t)$

$\vdots$

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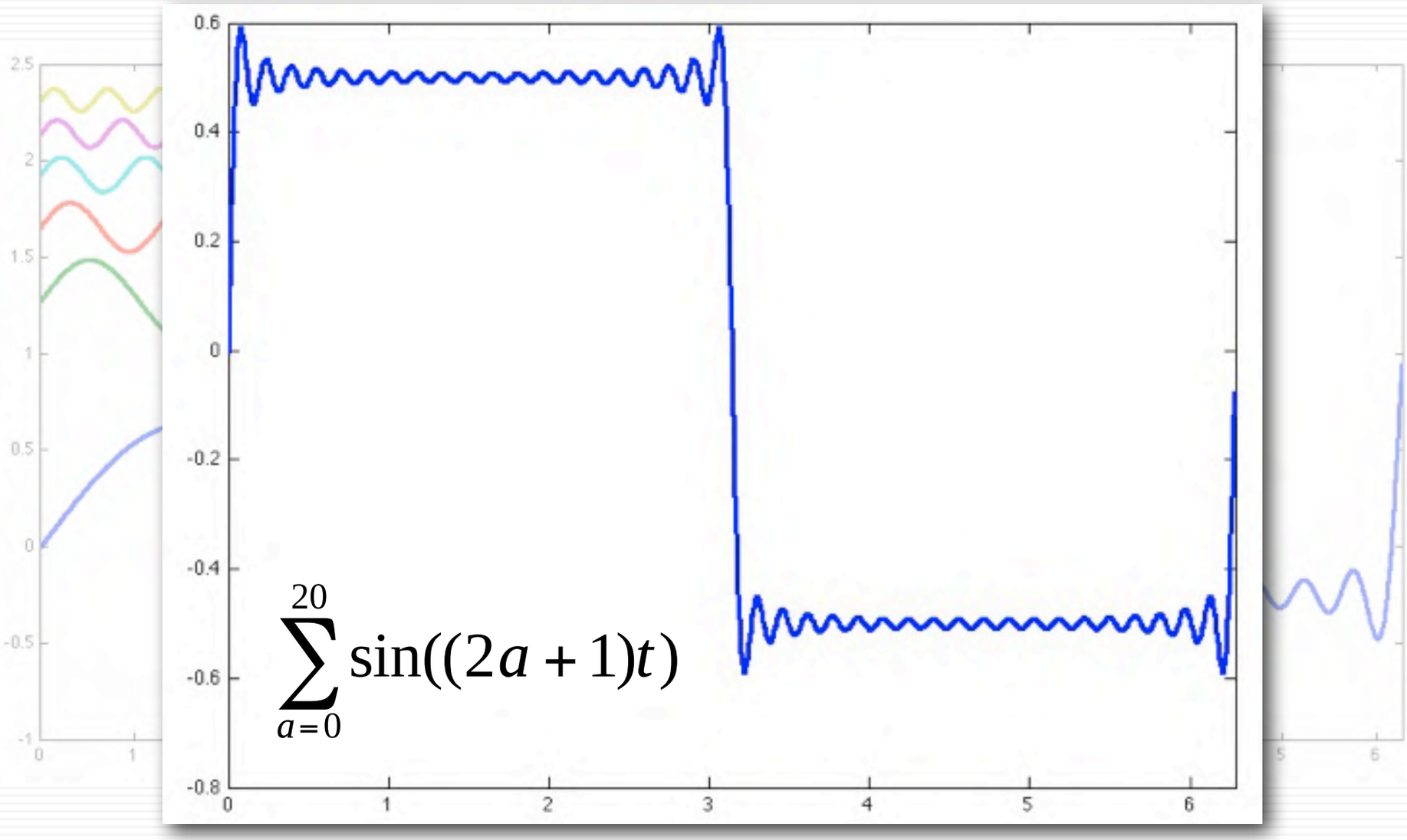
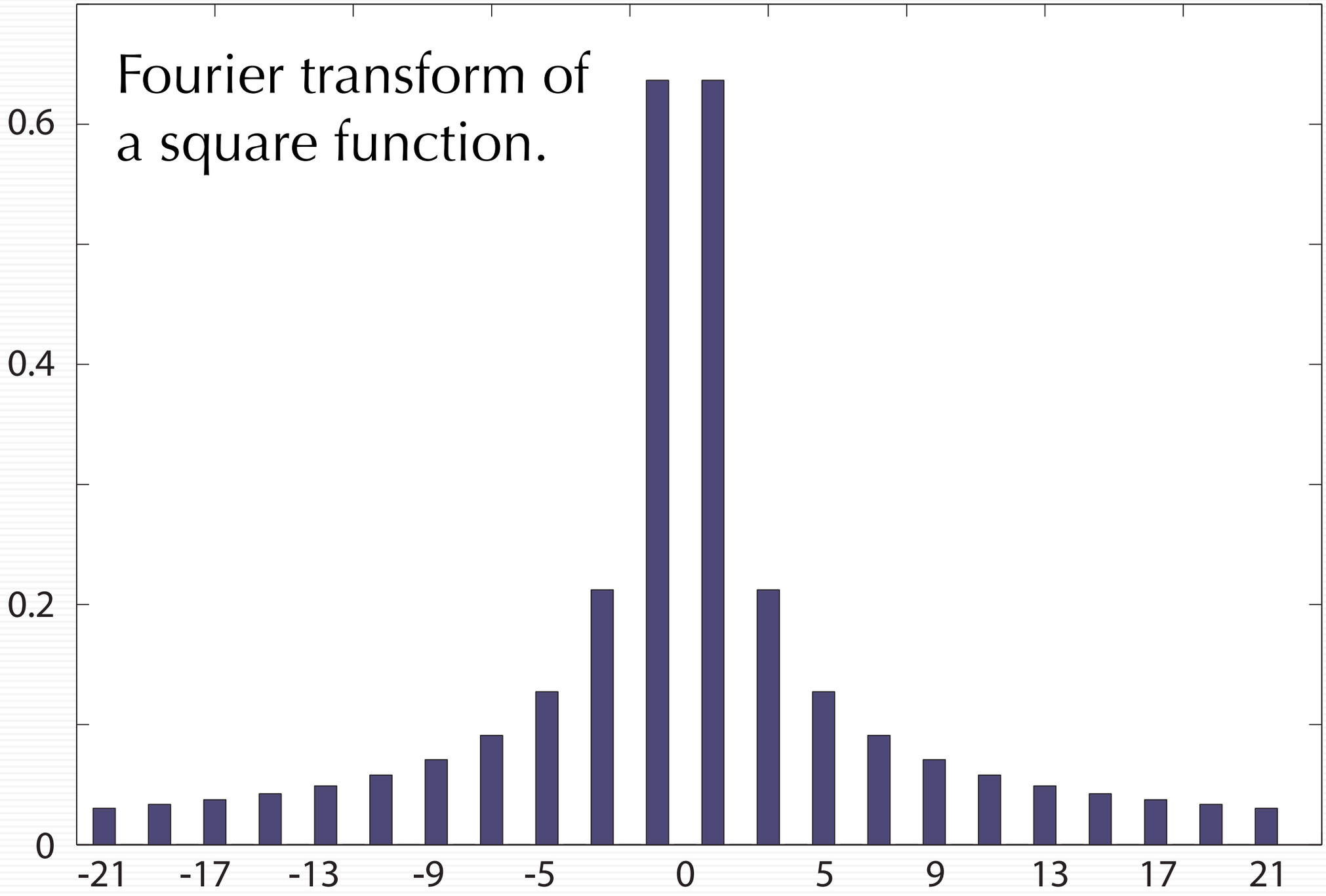
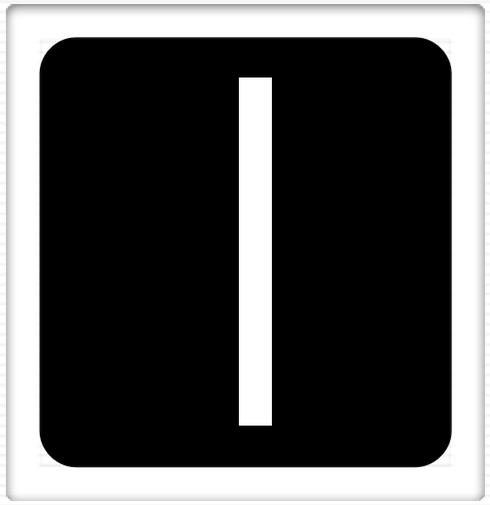


Typical Representation of Fourier Transform

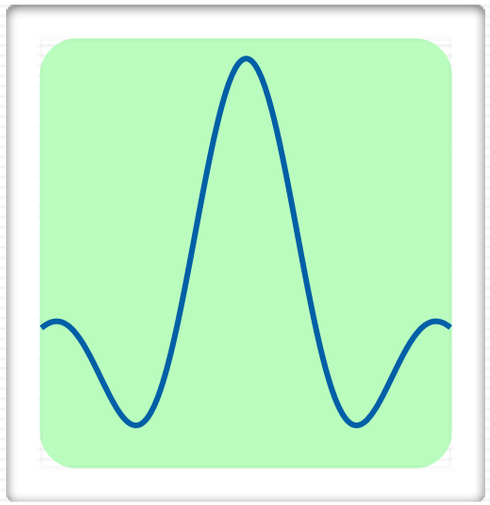


Objects with Finite Width

$f(t)$

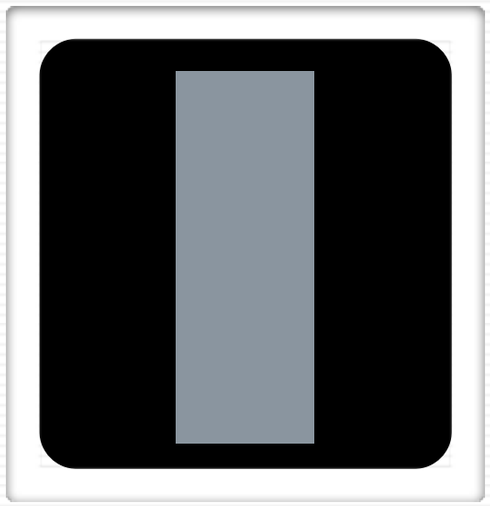


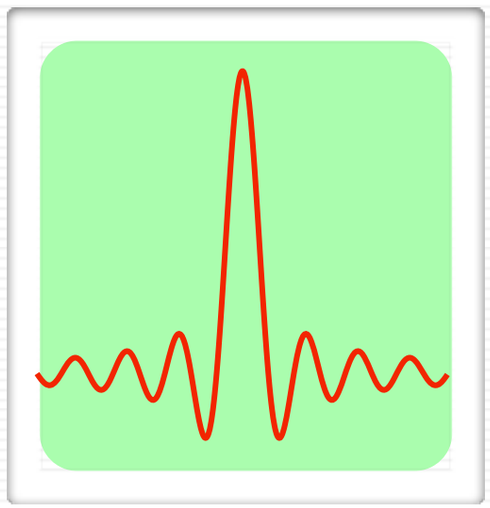
$F(\omega)$



Narrow Feature

➡ Wide Frequency Band

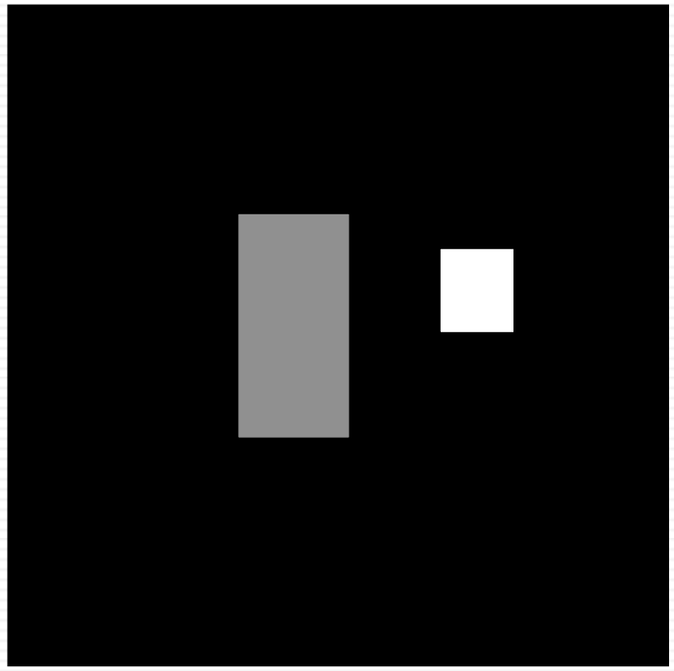




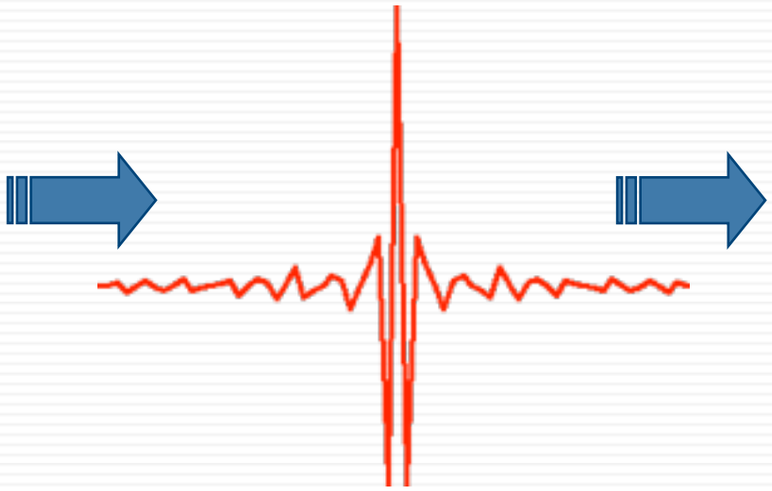
Wide Feature

➡ Narrow Frequency Band

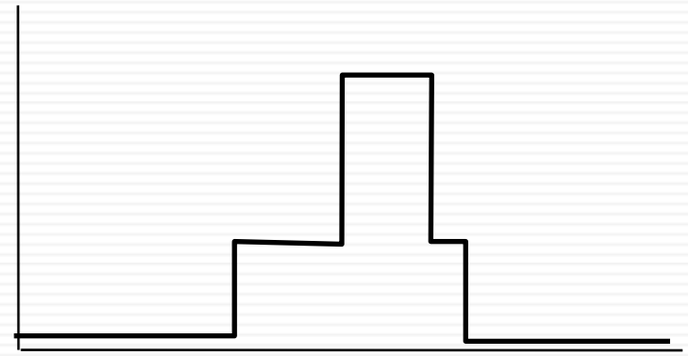
Fourier Projections



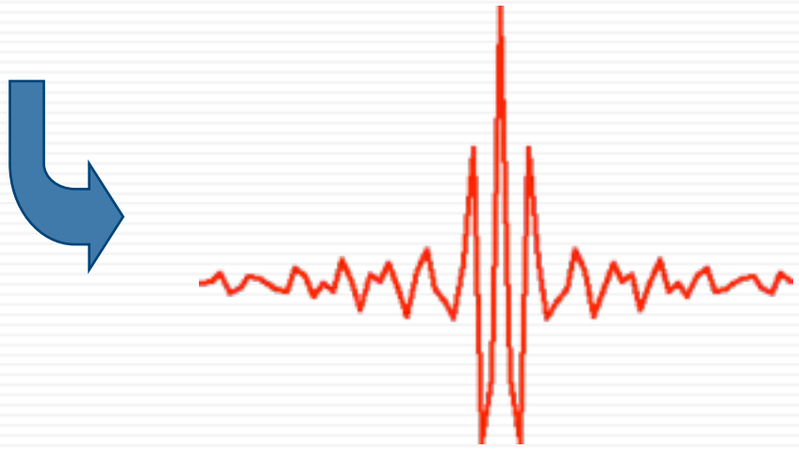
MR Image

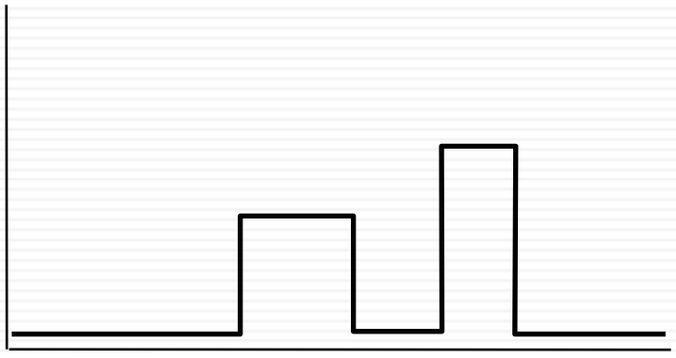


Raw Data



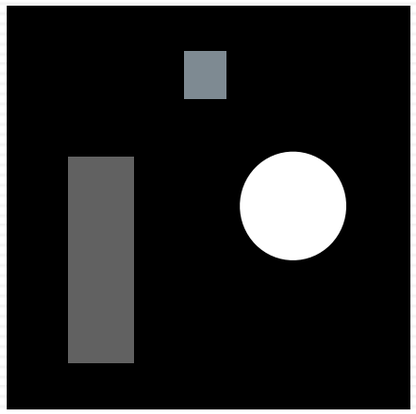
FFT of Raw Data



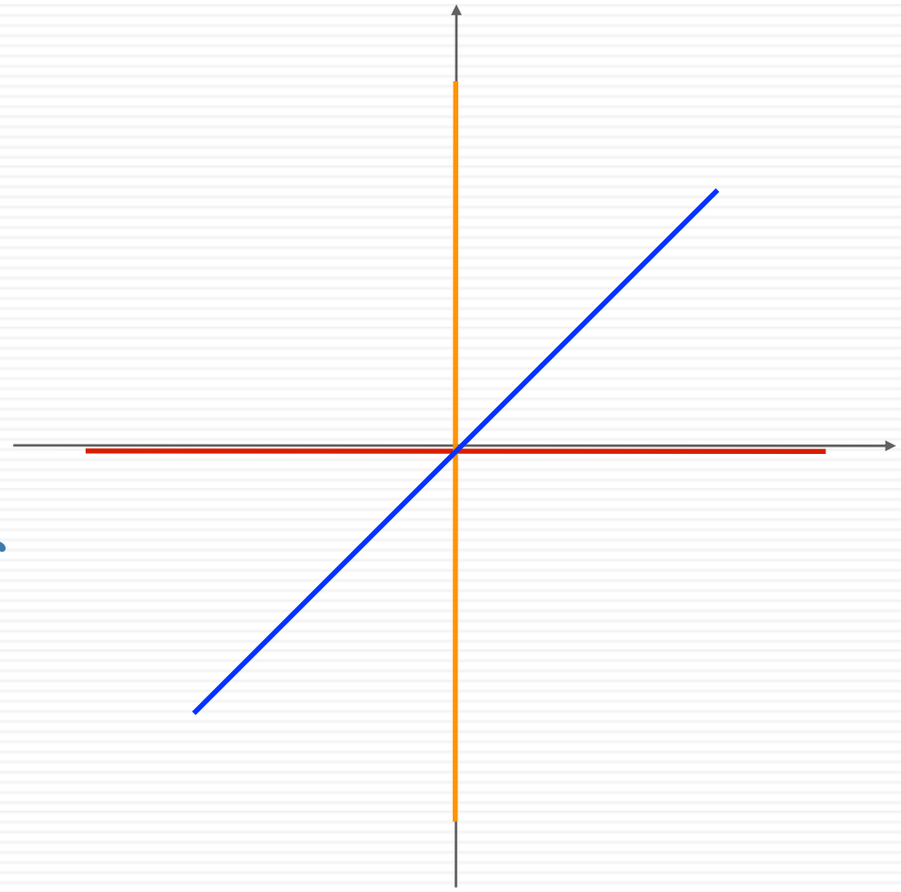


Back Projection

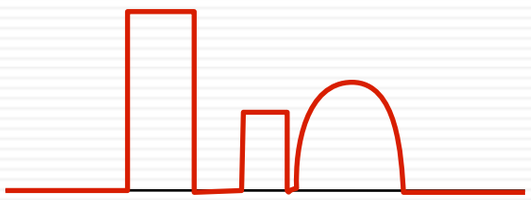
Image Domain

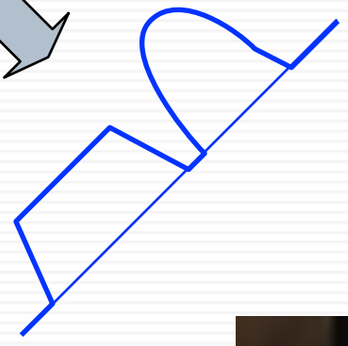


Gradient Encoding



2D Fourier Transform



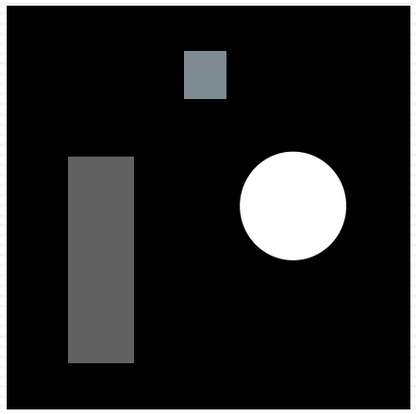


Fourier Domain

Paul Lauterbur

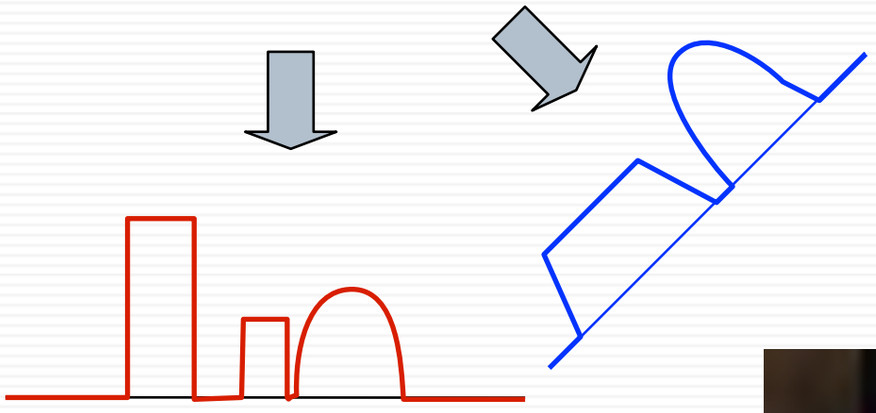
Back Projection

Image Domain



Gradient Encoding

2D Fourier Transform

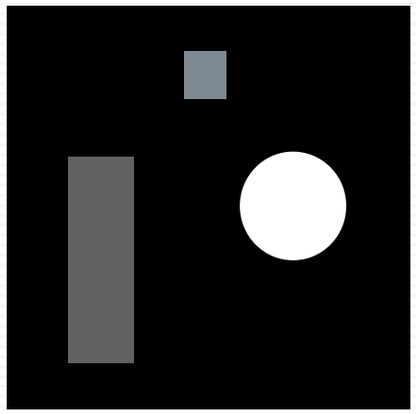


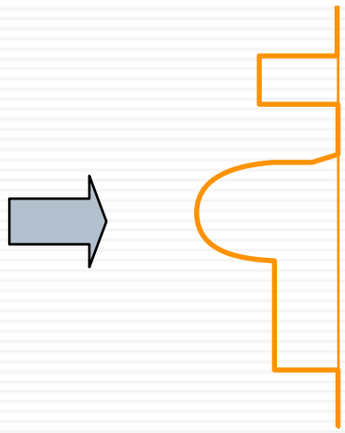
Fourier Domain

Paul Lauterbur

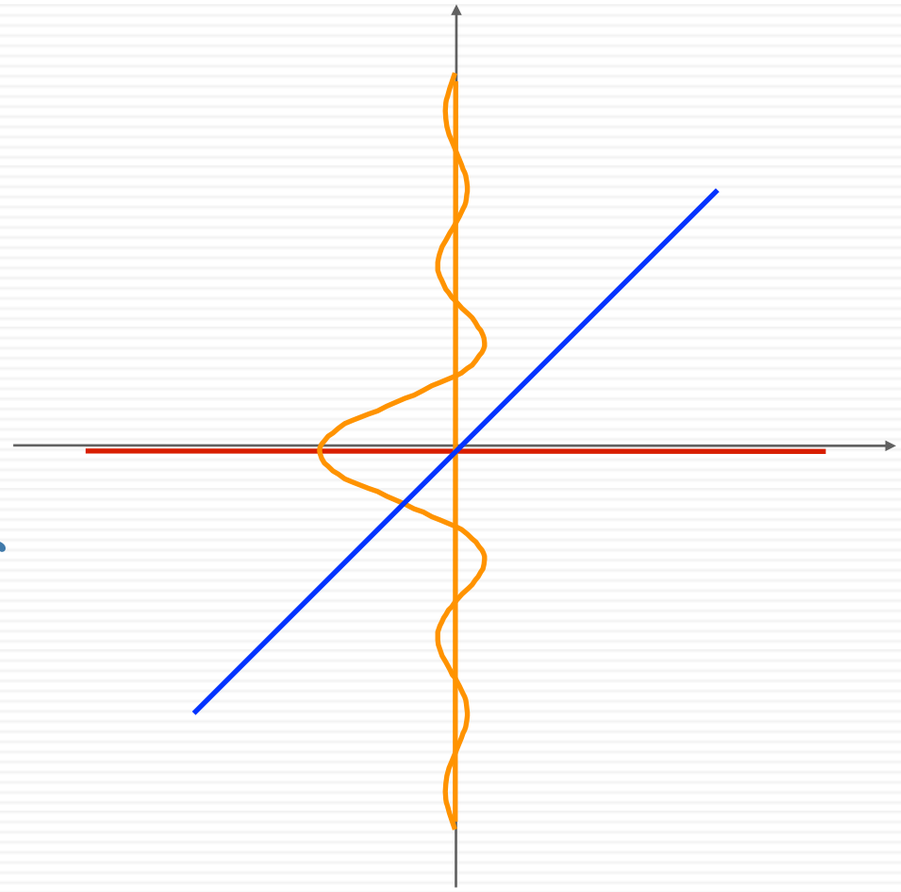
Back Projection

Image Domain





Gradient Encoding



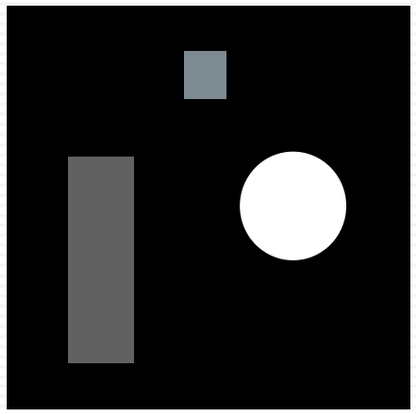
2D Fourier Transform

Fourier Domain

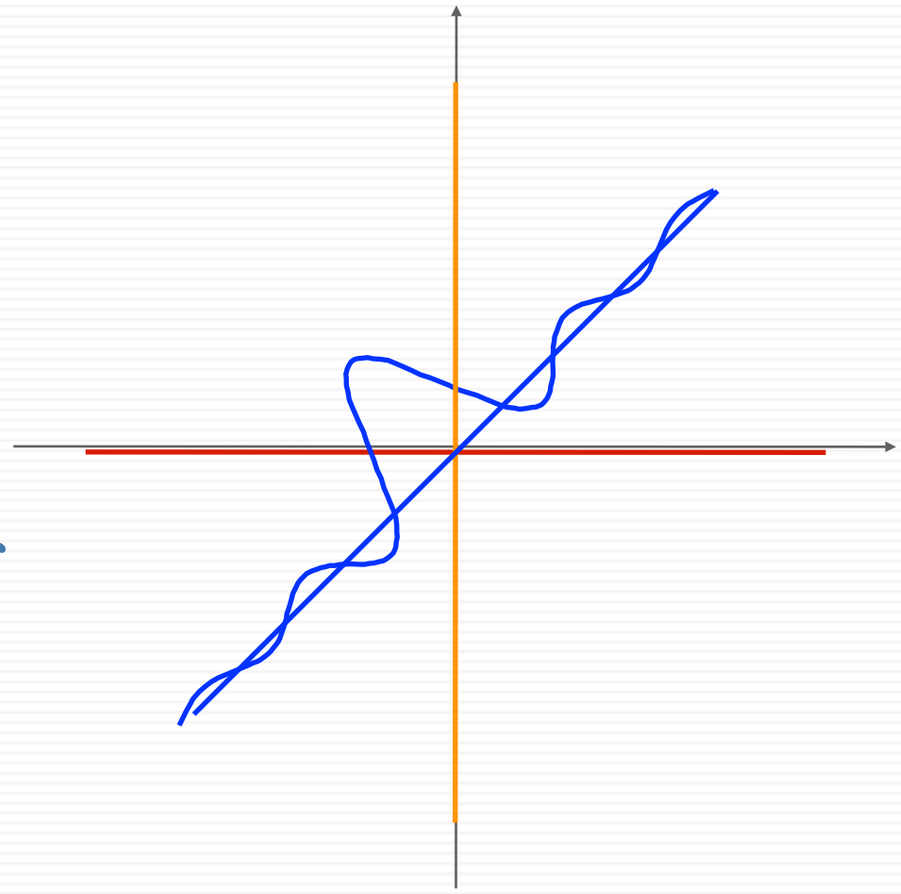
Paul Lauterbur

Back Projection

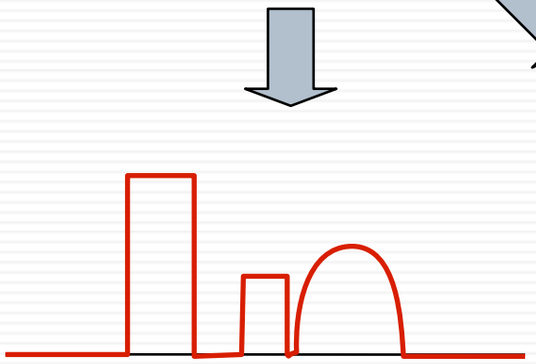
Image Domain

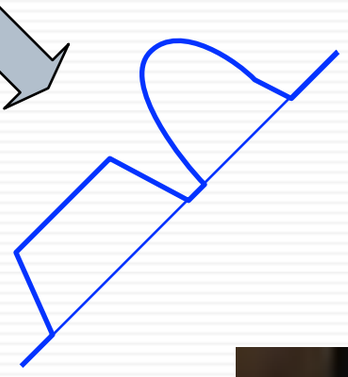


Gradient Encoding



2D Fourier Transform



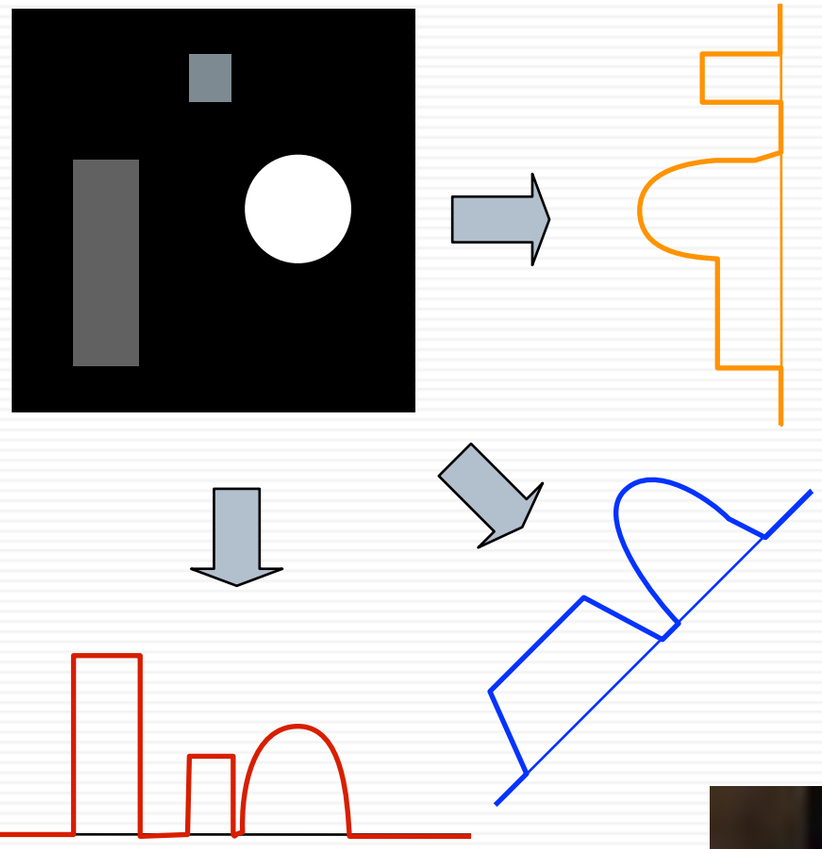


Fourier Domain

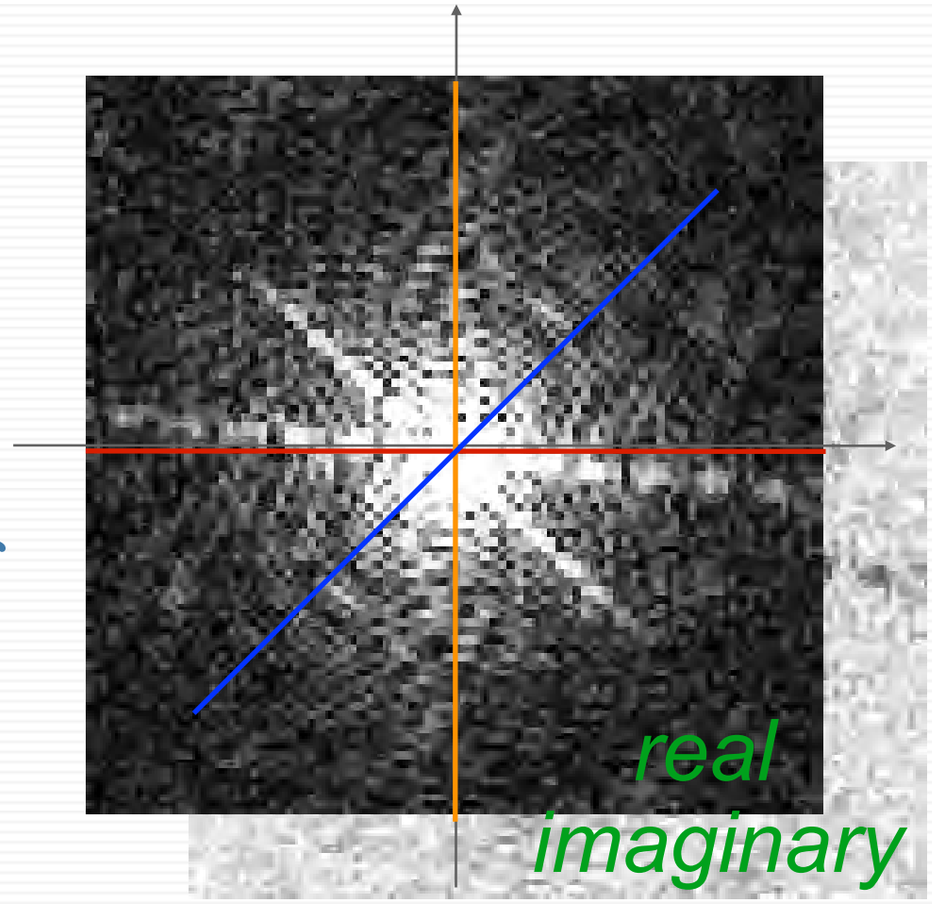
Paul Lauterbur

Back Projection

Image Domain



Gradient Encoding

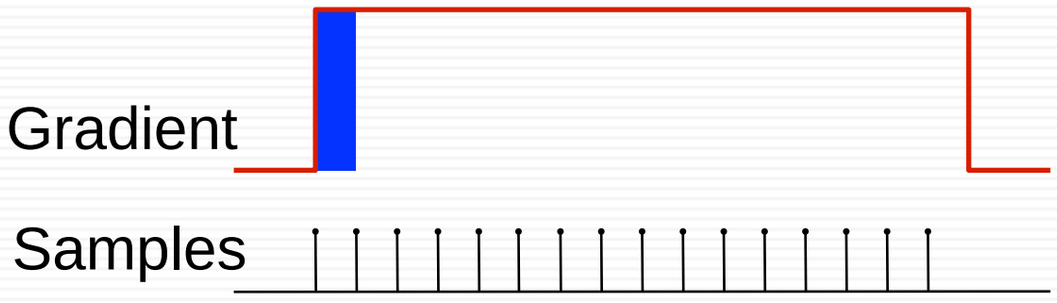


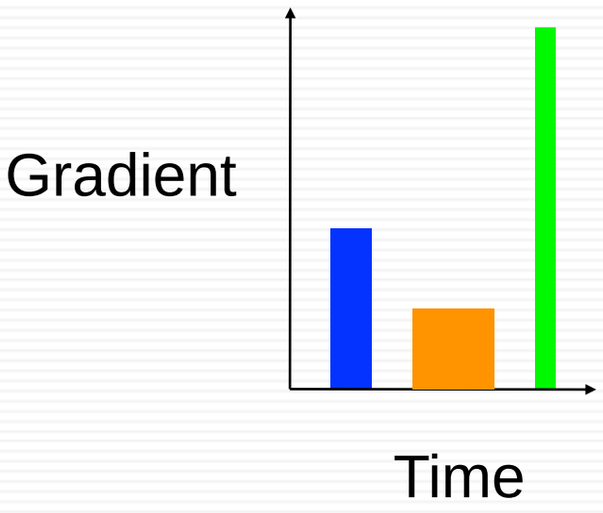
2D Fourier Transform

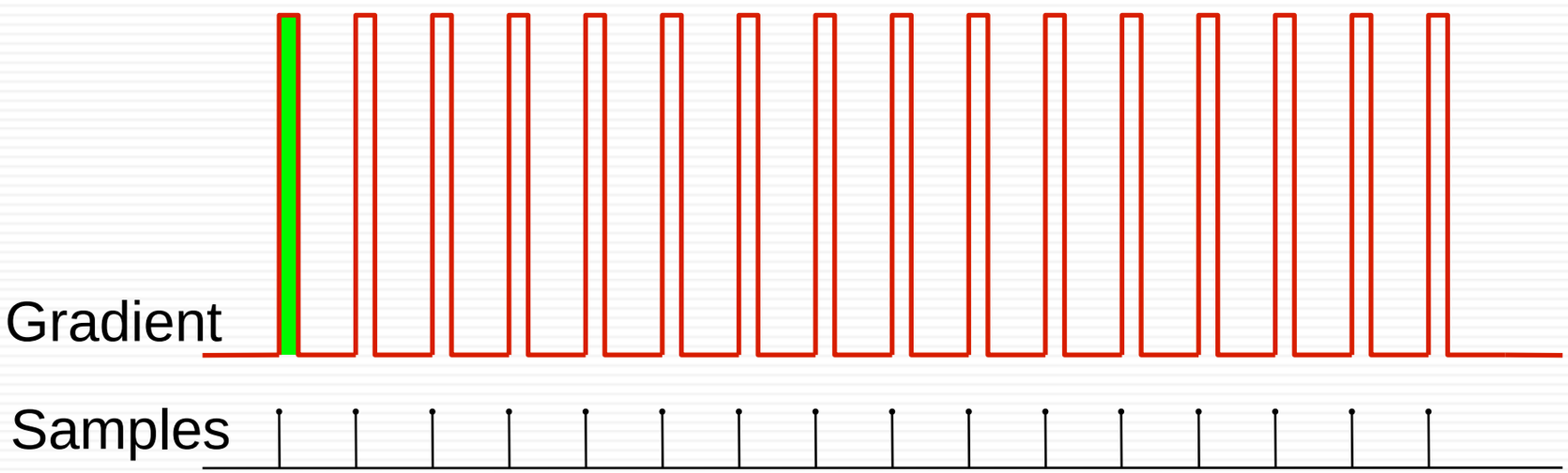
Fourier Domain

Paul Lauterbur

Equivalent Strategies in k-space*

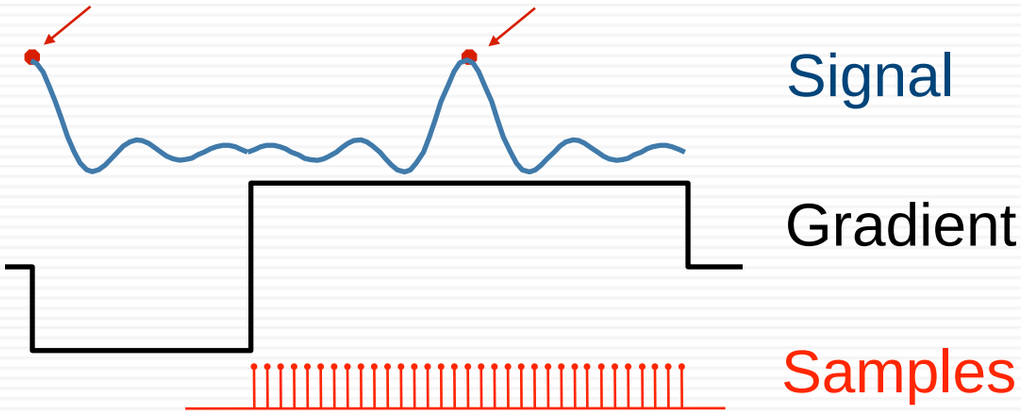






**Ignoring effects of signal decay and sample motion*

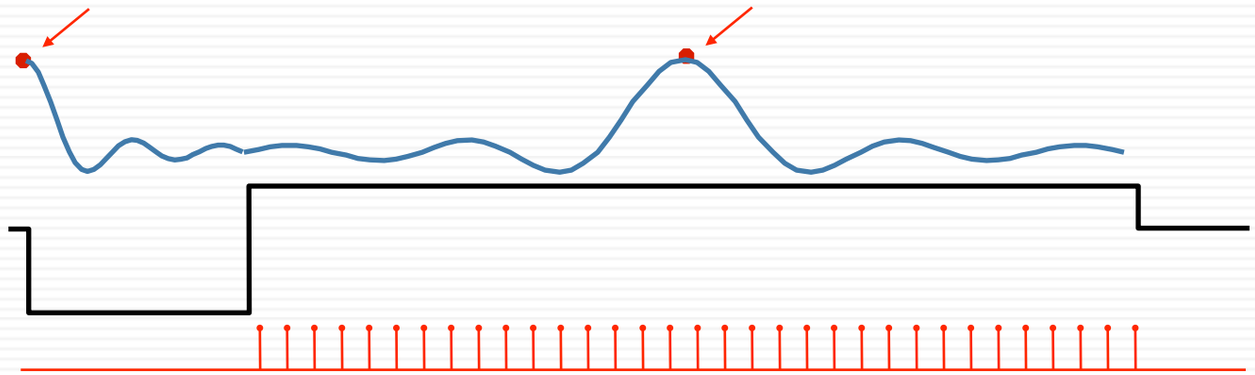
Gradient Pre-encoding



Signal

Gradient

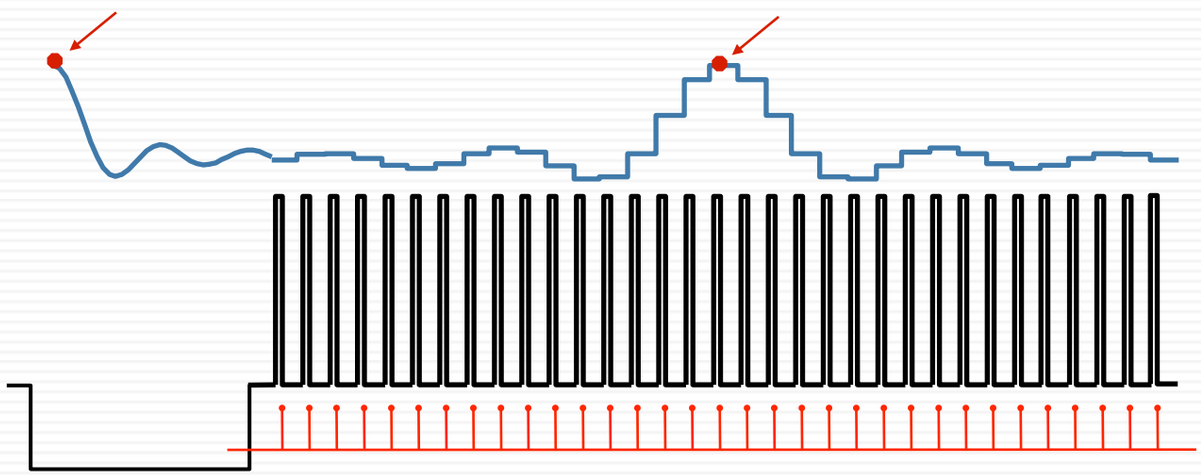
Samples



Signal

Gradient

Samples



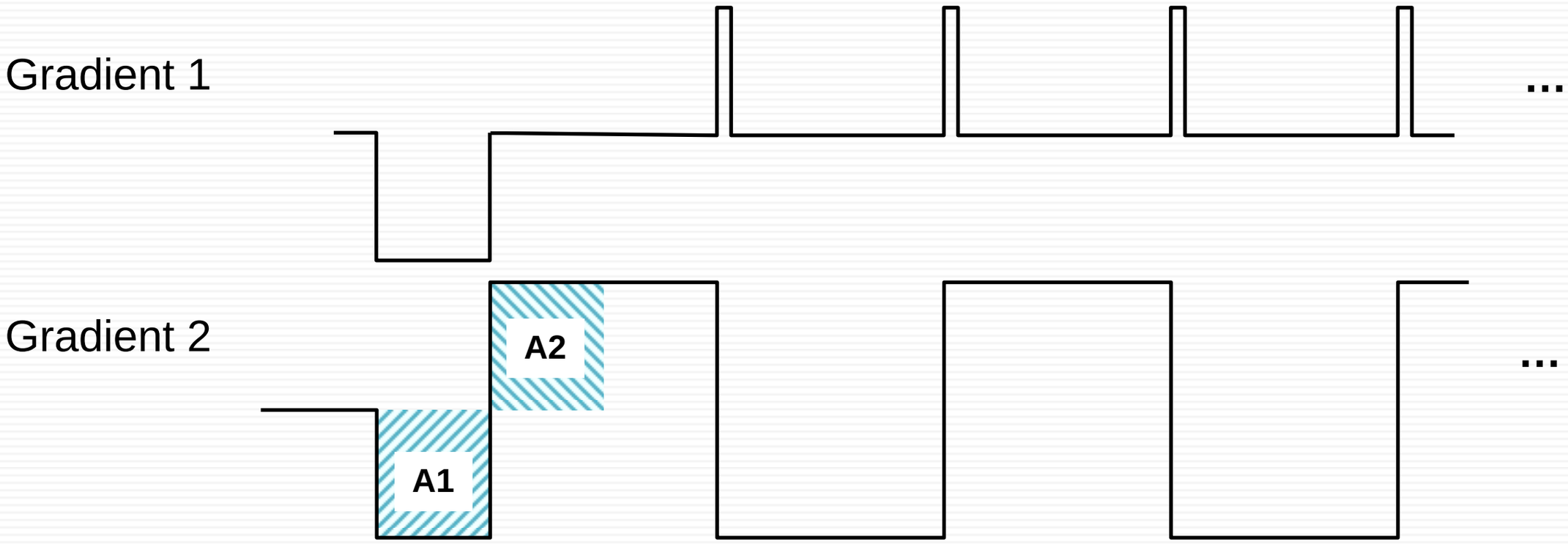
Time

Signal

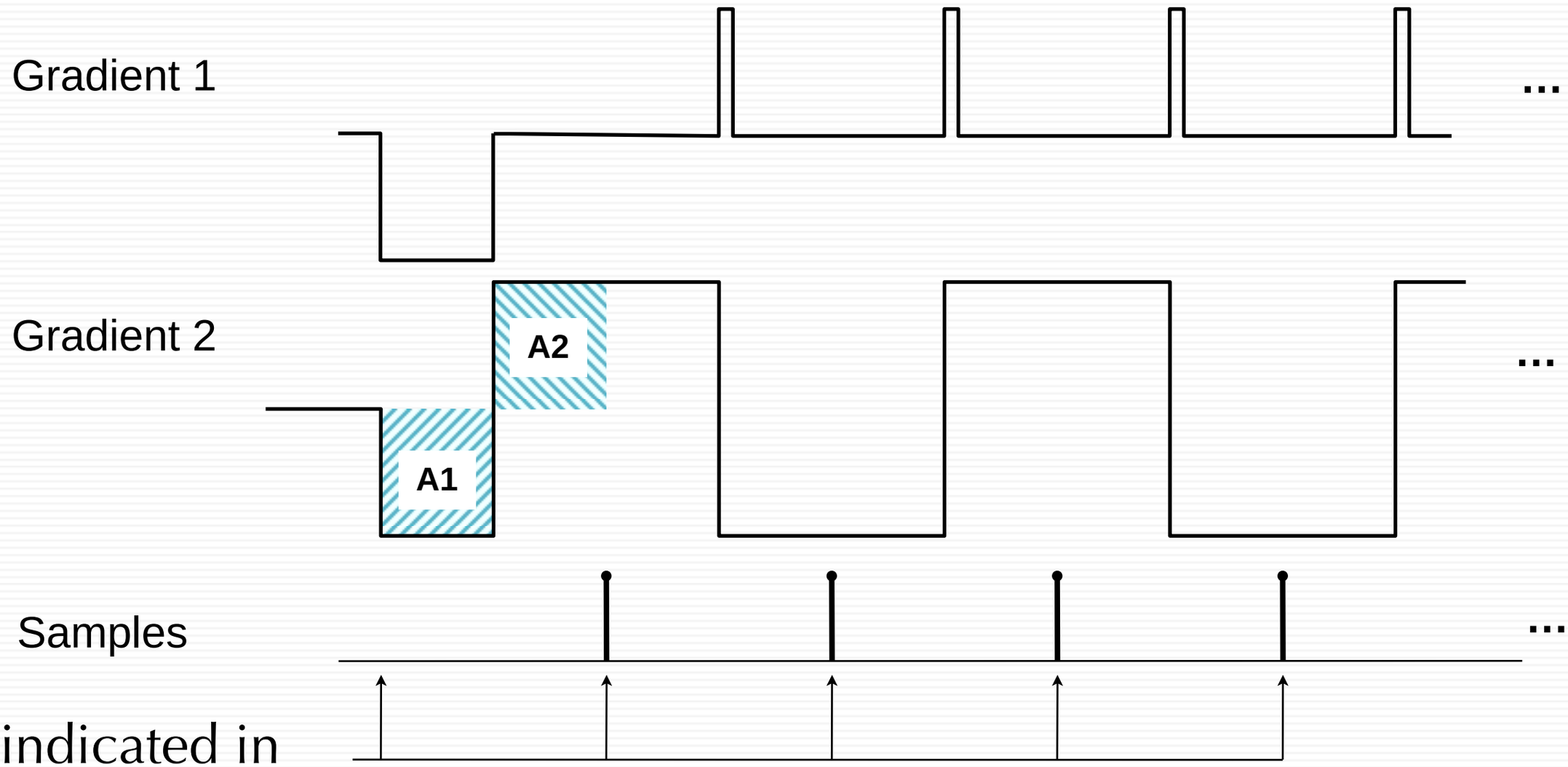
Gradient

Samples

Interleaved Spatial Encoding

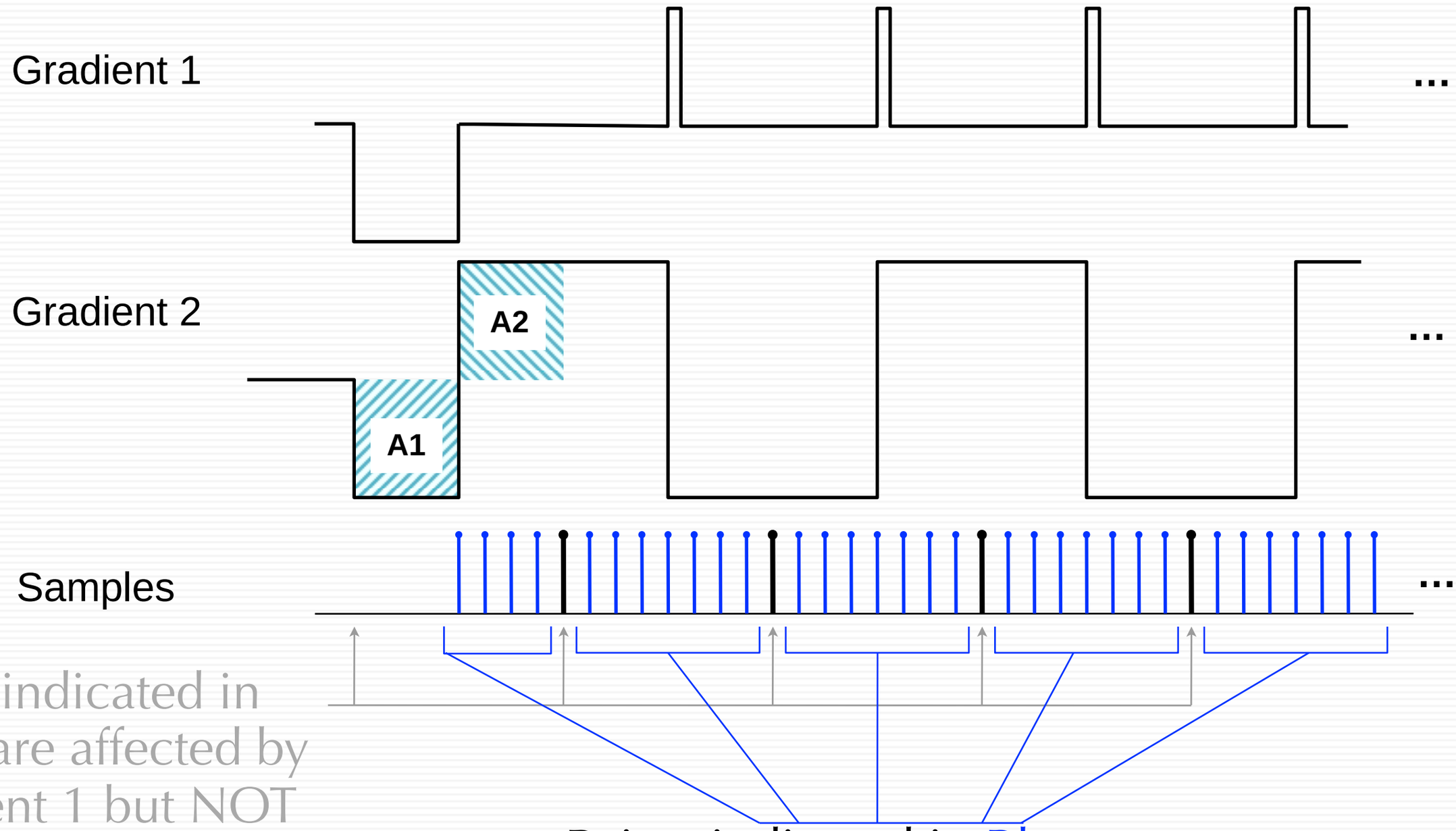


Interleaved Spatial Encoding



Points indicated in black are affected by Gradient 1 but NOT by Gradient 2

Interleaved Spatial Encoding



Points indicated in black are affected by Gradient 1 but NOT by Gradient 2

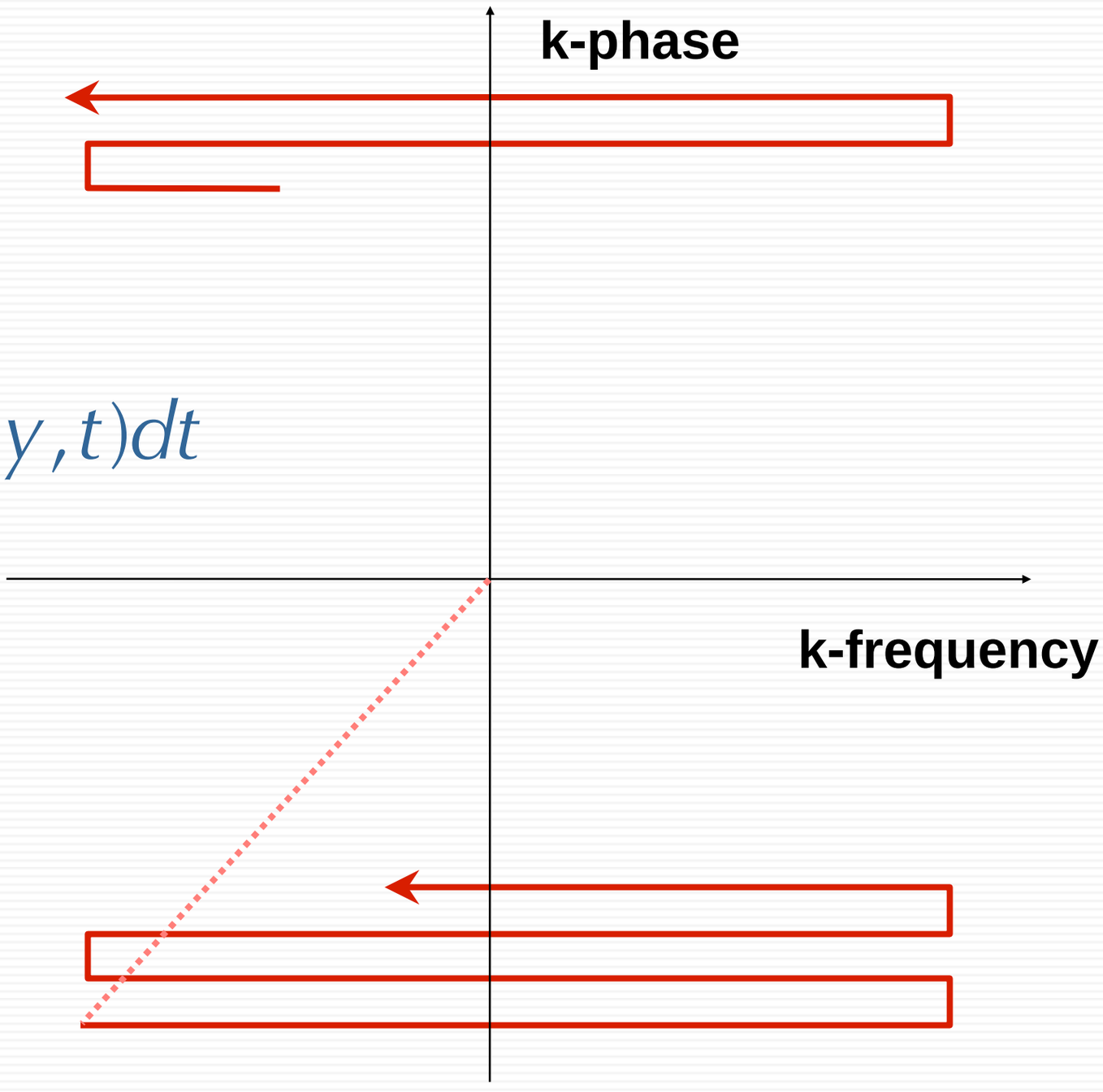
Points indicated in Blue are affected by both gradients

Echo-Planar k-space Trajectory

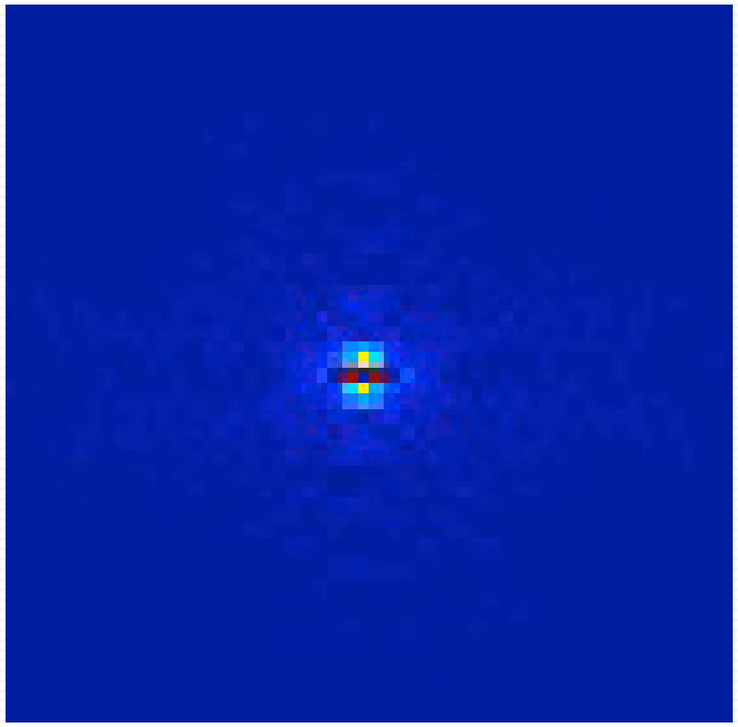
k-space plots the integral of the gradient encoding.

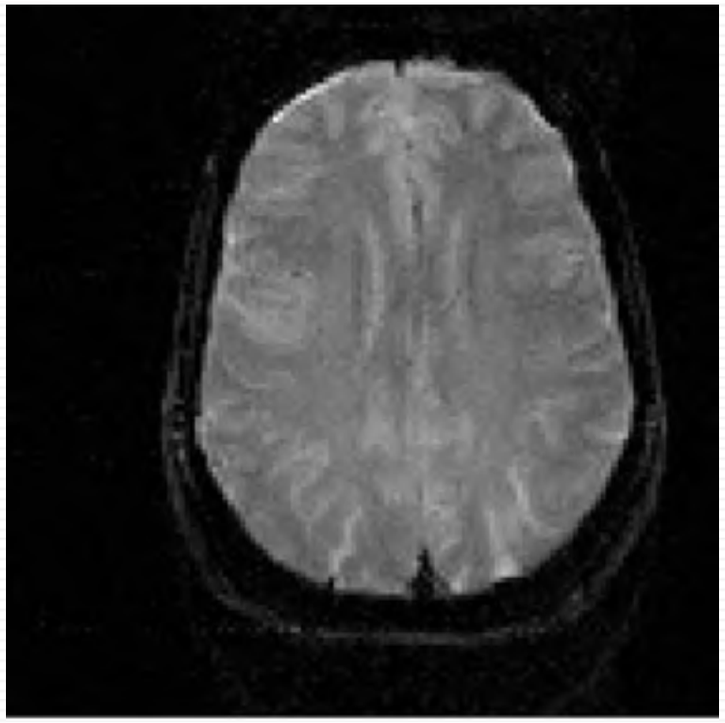
$$k(x,y,t) = \gamma \int_0^T G(x,y,t)dt$$

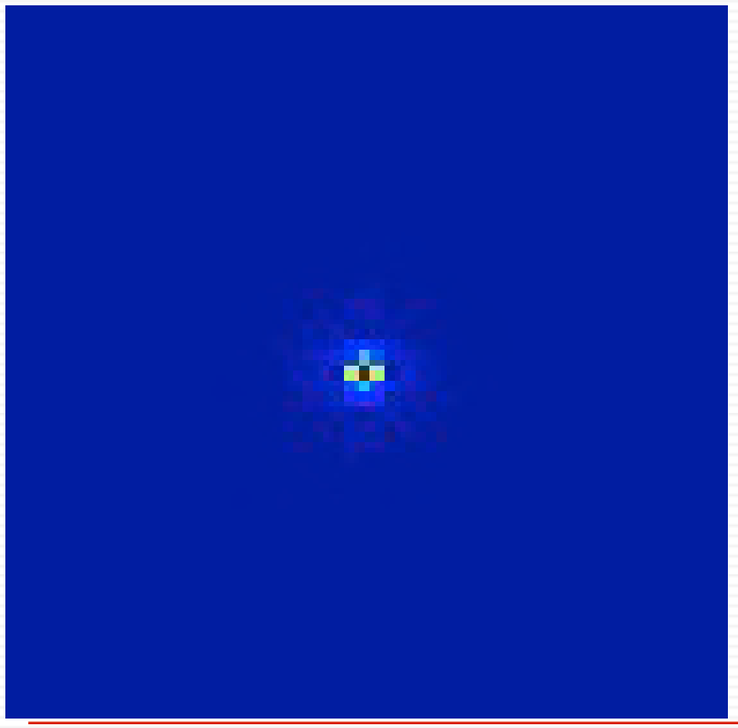
Its Fourier transform is the image.

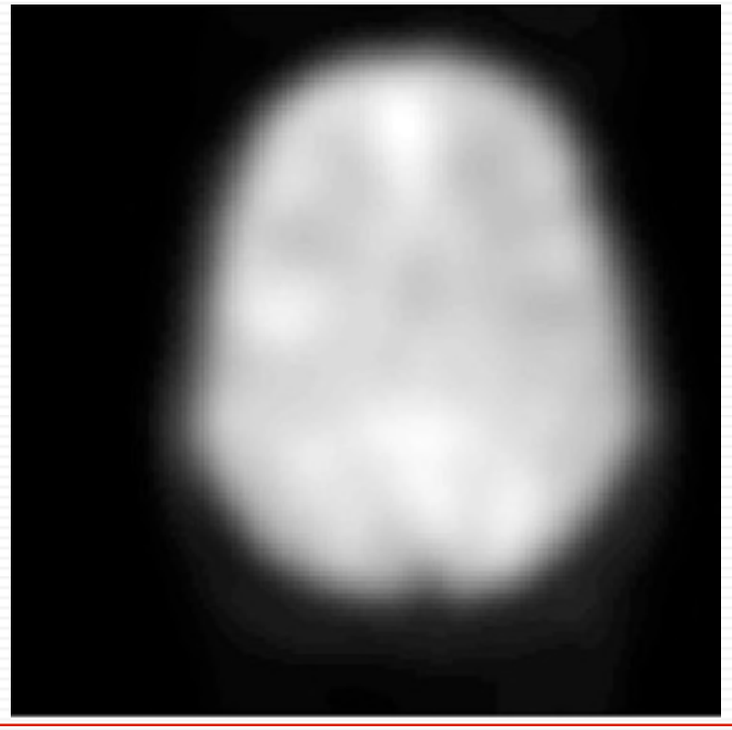


Properties of K-Space



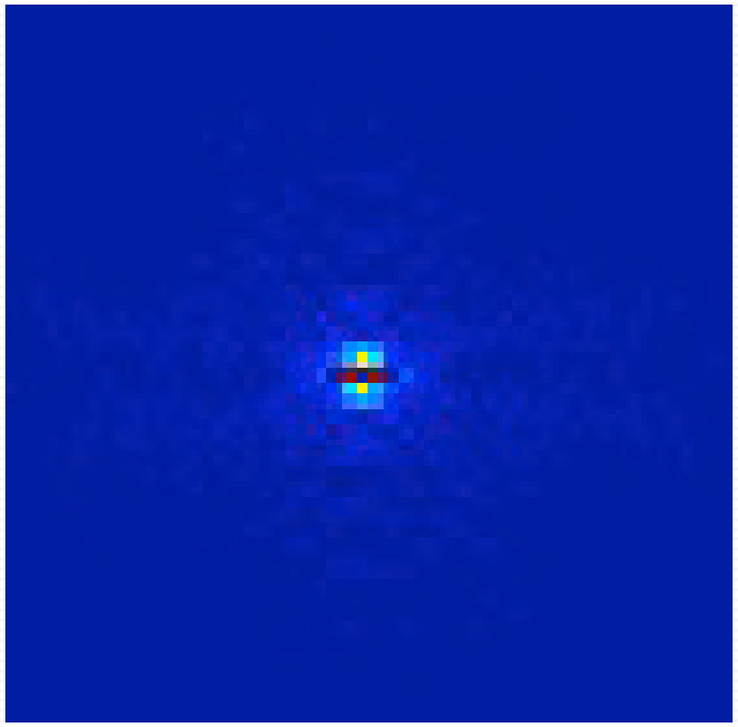


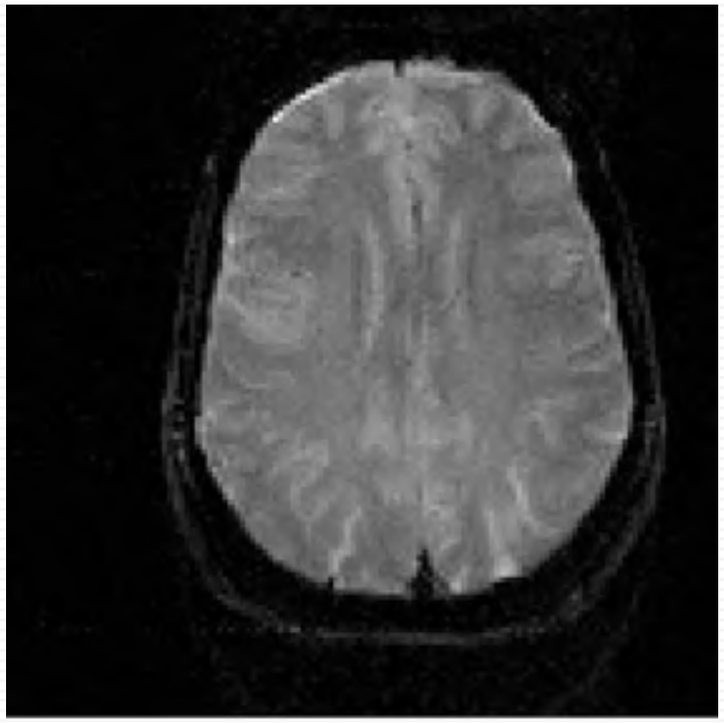


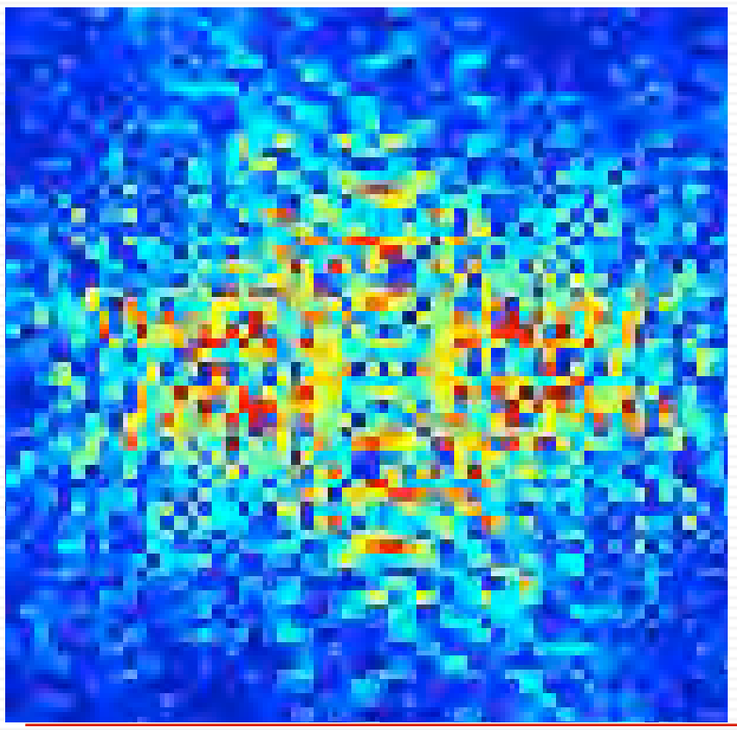


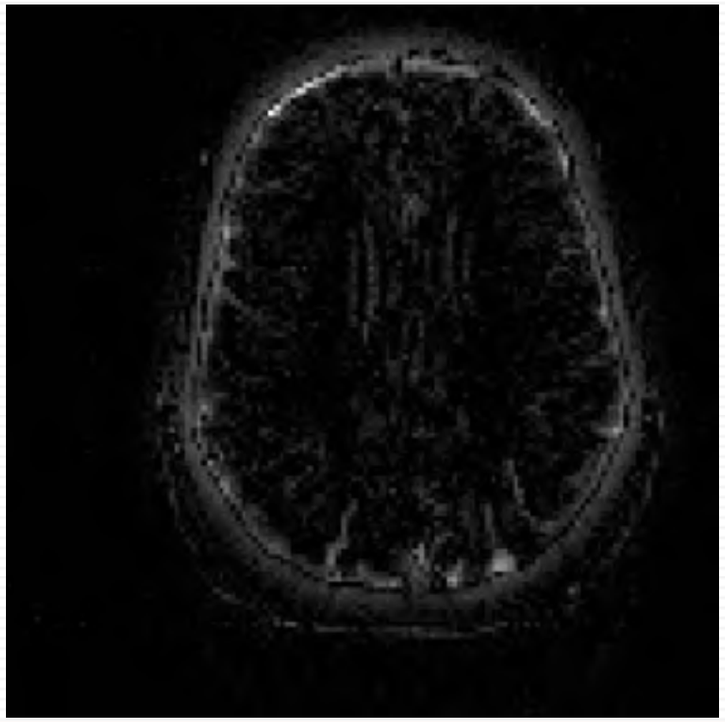
- Increasing K values Represent Higher Spatial Frequencies, thus Higher Resolution
- Finer Grain Sampling Results in Wider FOV

Properties of K-Space









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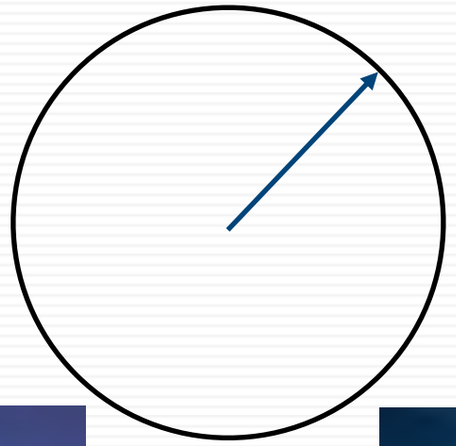
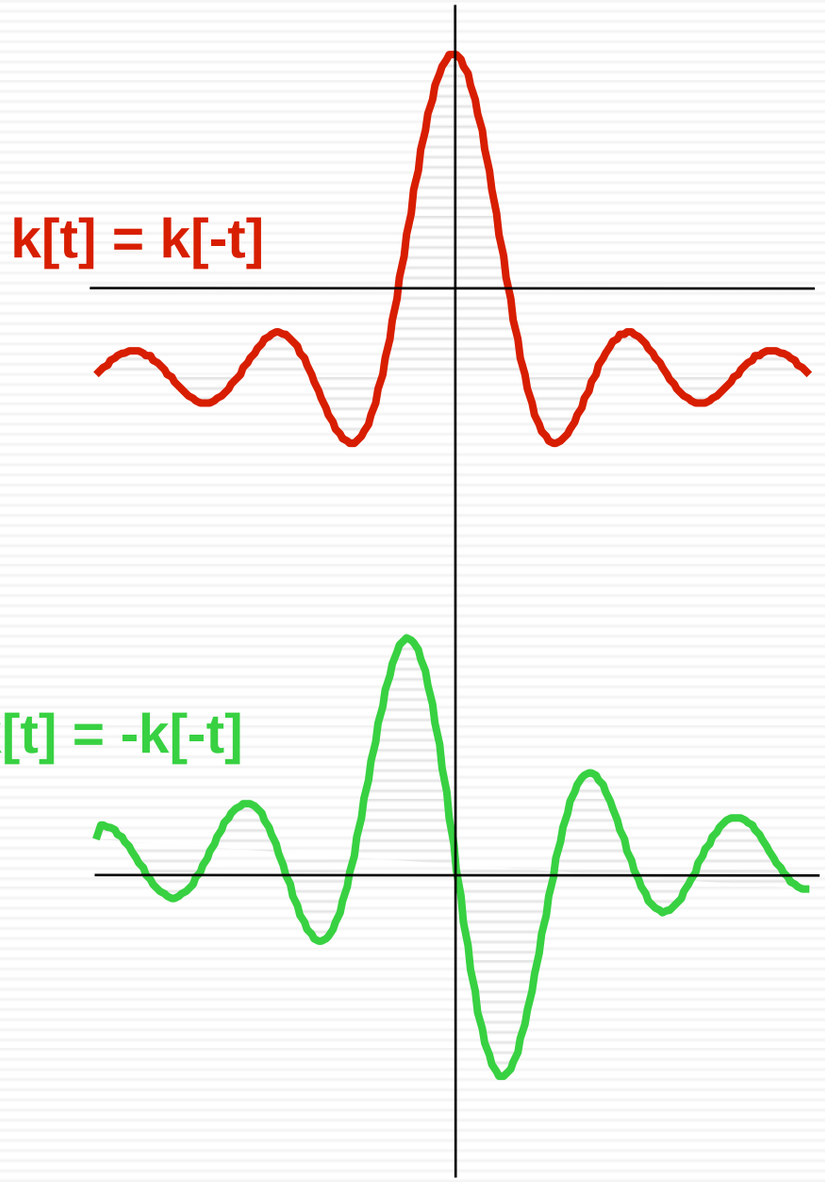
Raw Data Symmetry



Detector 1

Detector 2



k-space conjugate symmetry

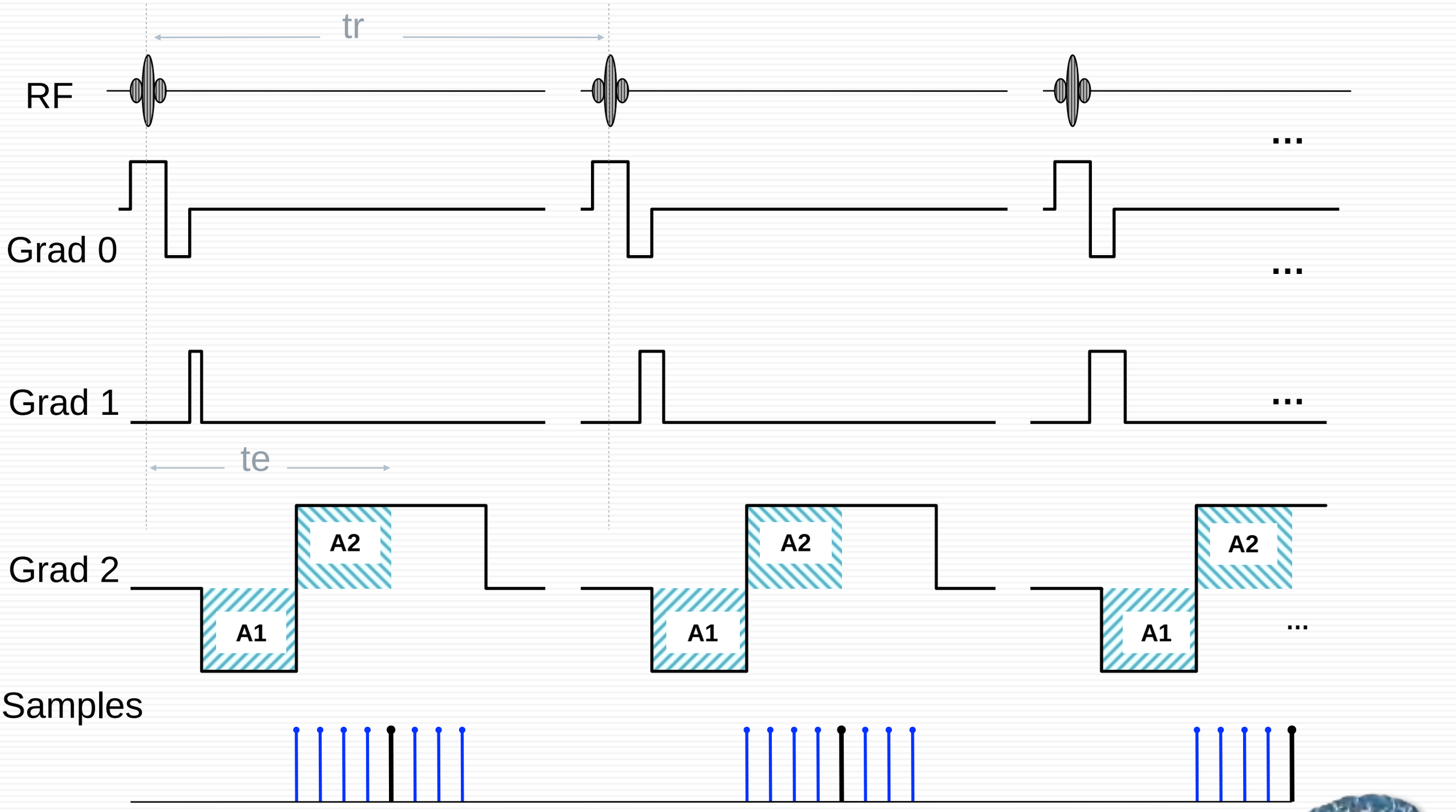
For a Stationary Object,
in a Homogeneous Field: $S(kx,ky) = \overline{S(kx,-ky)}$

where $S(kx, ky)$ is the signal at (kx,ky).

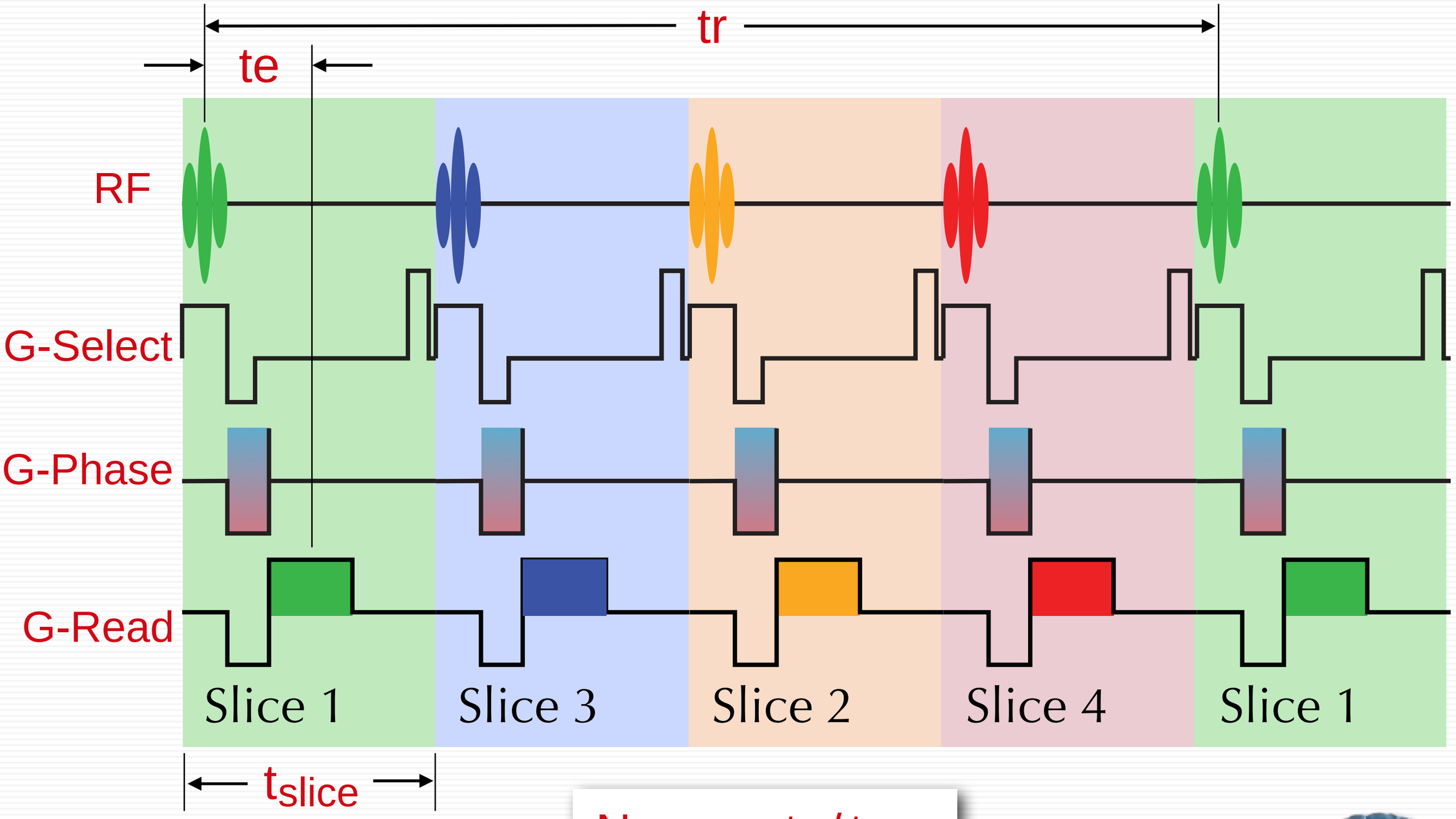
Example: if $S(kx,ky) = \overline{a+ib}$,

then $S(kx,-ky) = \overline{(a+ib)} = a-ib$.

Conventional Spatial Encoding

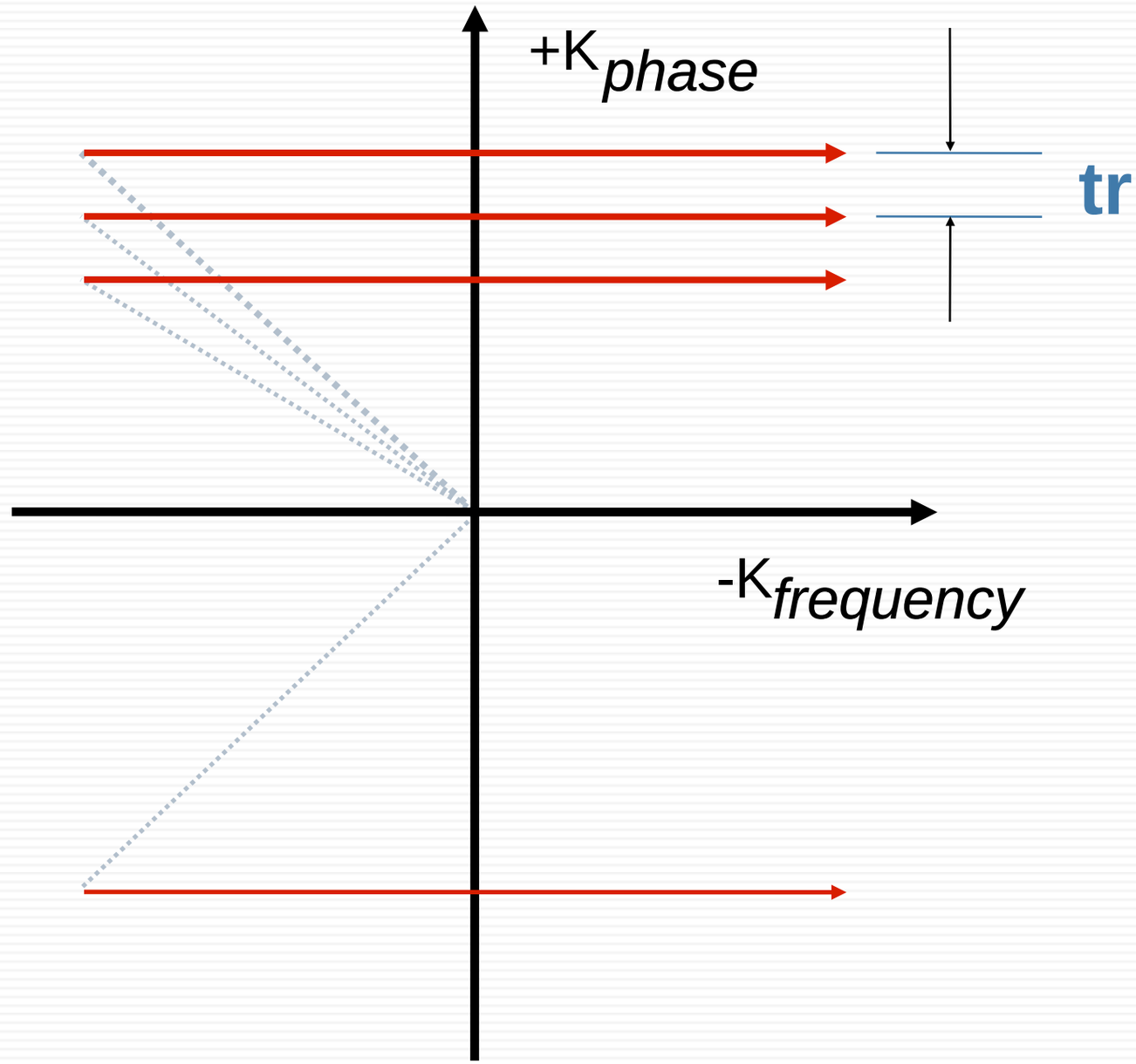


Multi-slice MRI

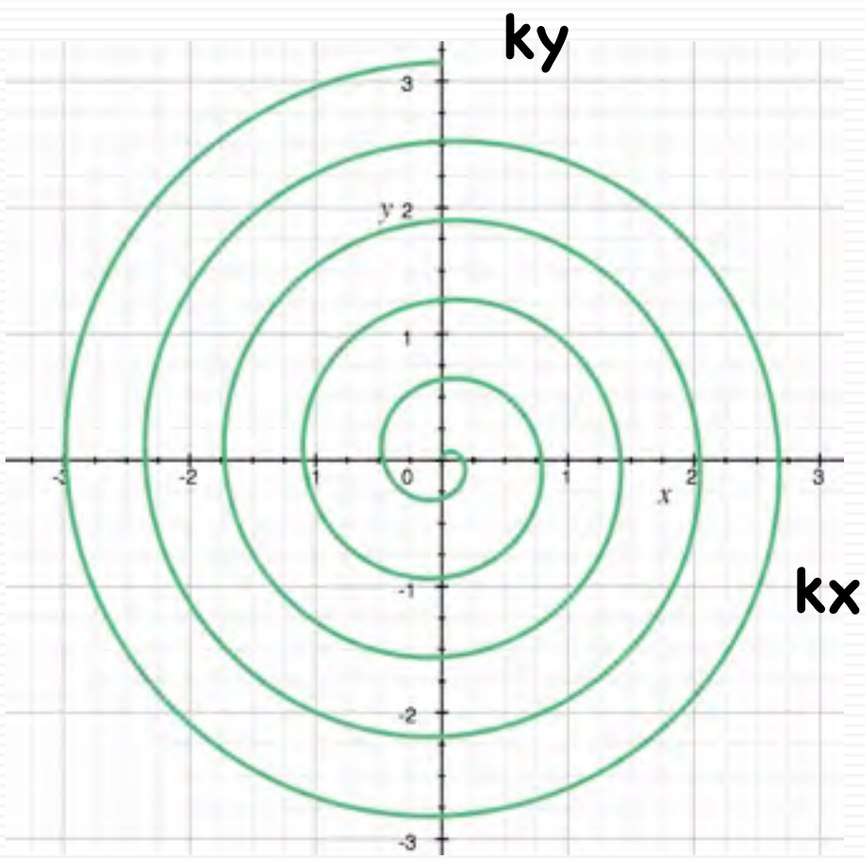


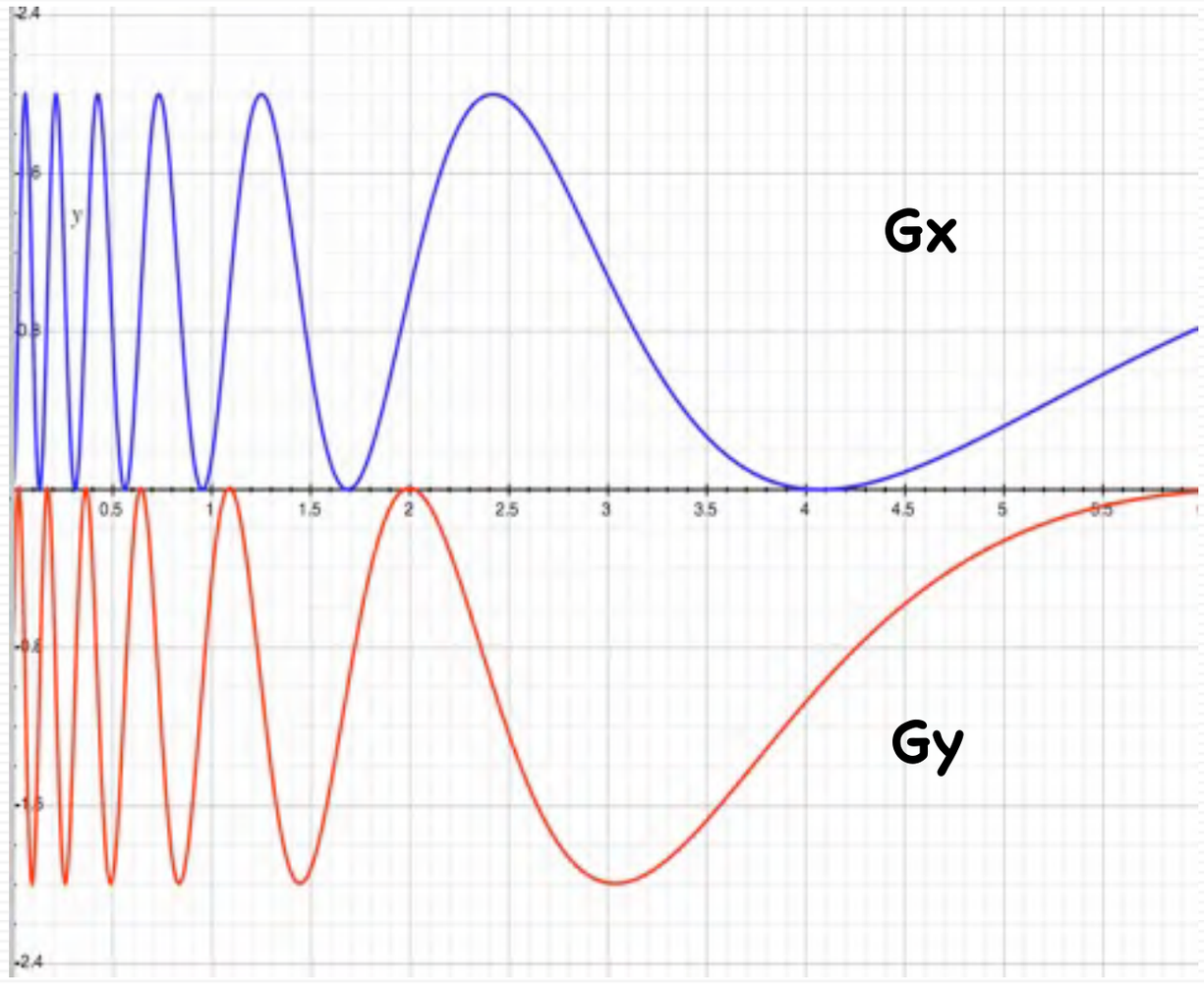
$$N_{slices} = tr / t_{slice}$$

Conventional K-Space Trajectory

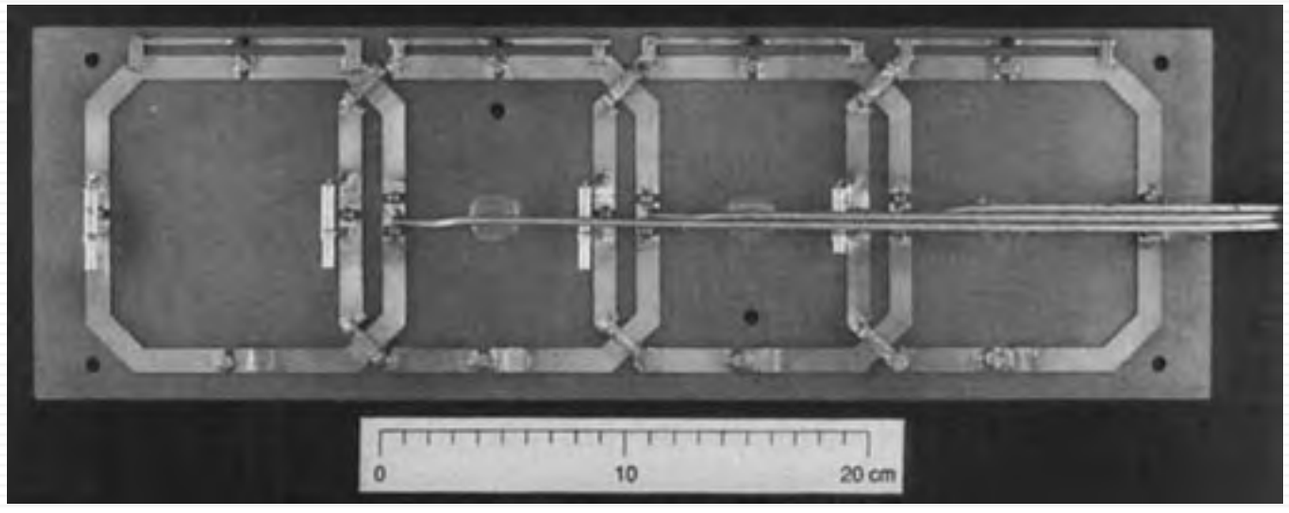


Spiral





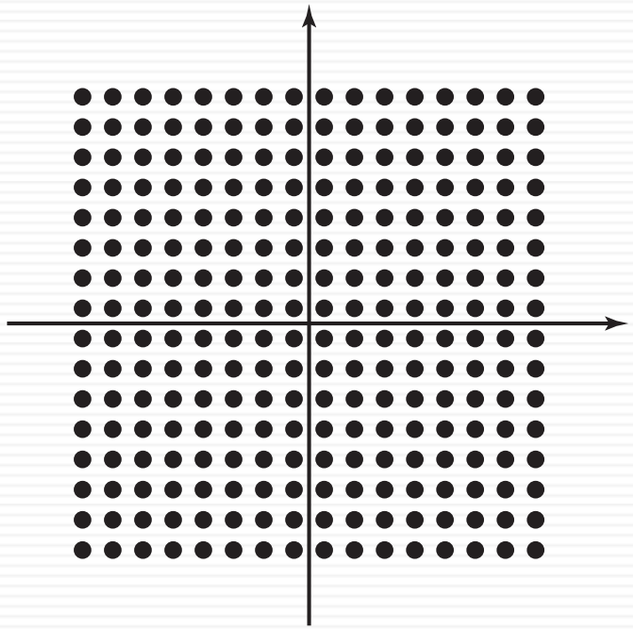
The NMR Phased Array

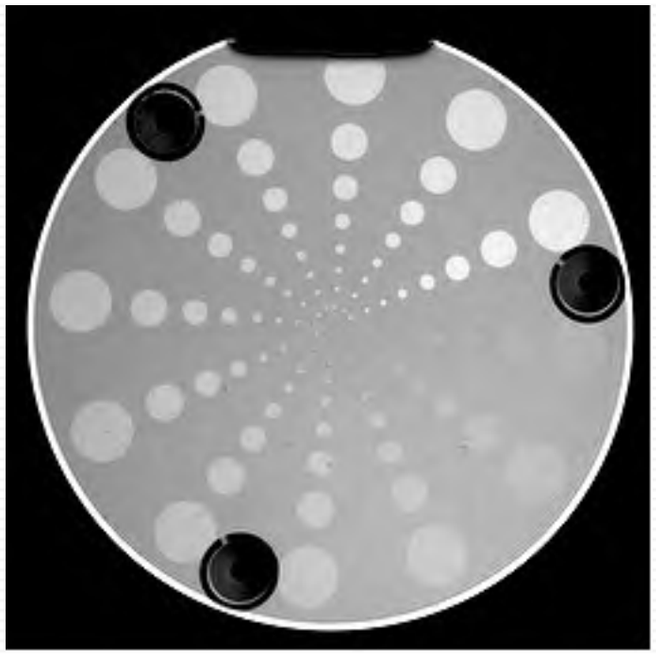


Peter Roemer. et al.,

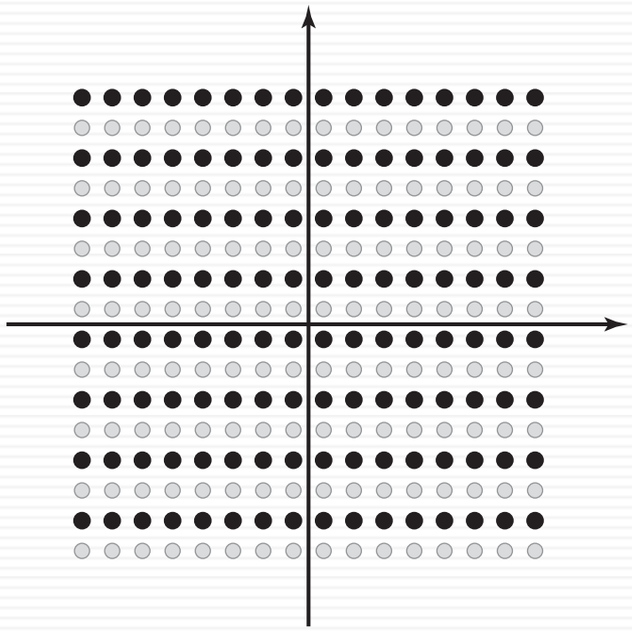
Magn Reson Med. 16:192, 1990

SENSE Encoding

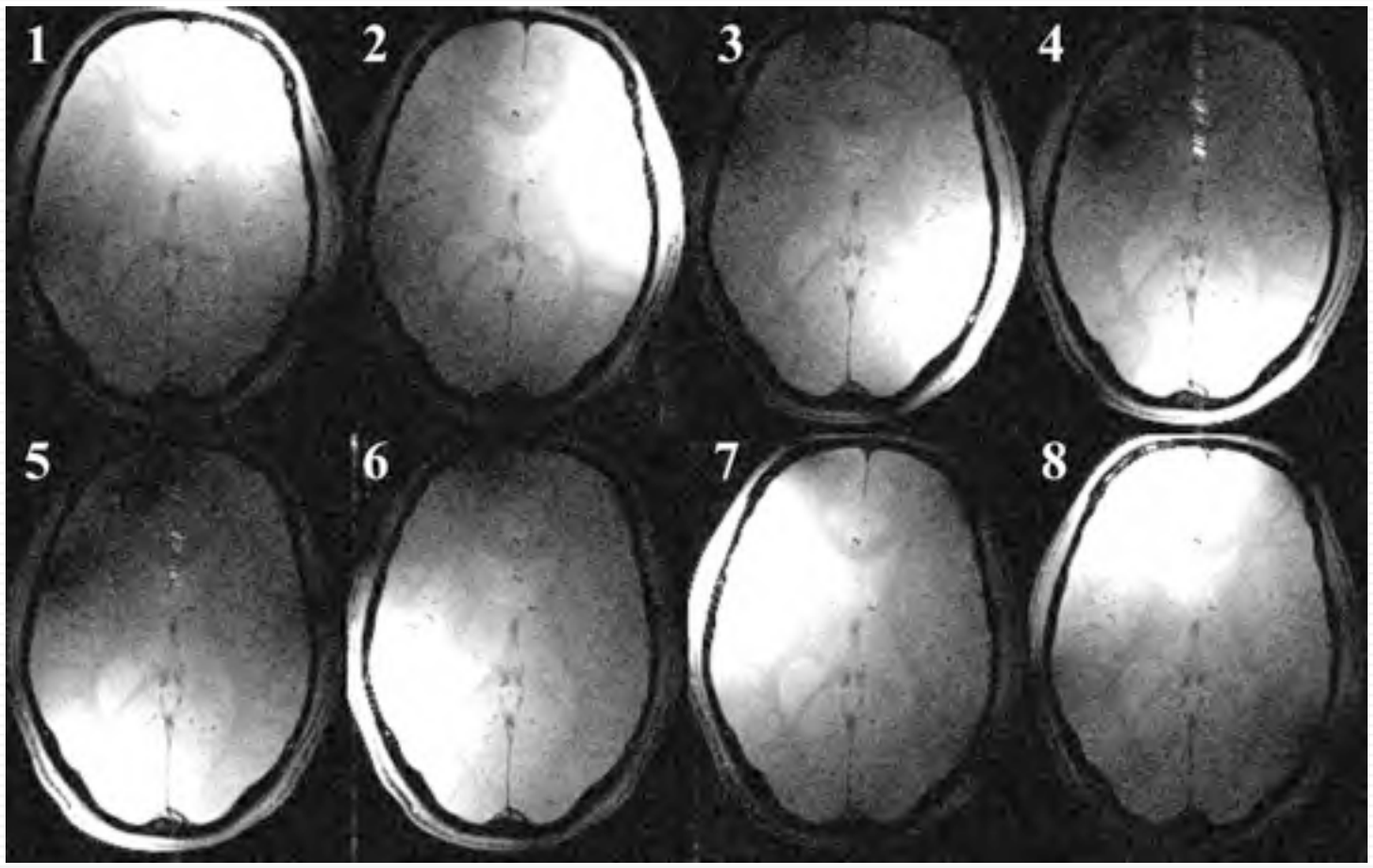




K. P. Pruessmann, et al.,
Magn Reson Med. 42:952, 1999



http://www.mr.ethz.ch/sense/



Spatial Encoding Summary

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- Hermite Symmetry of the Raw Data may be used to Reduce Scan Times;
- Literally Hundreds of Pulse Sequences are in Common Use,

The Plan

- The Magnetic Resonance Phenomenon & Contrast (30)
- Spatial Encoding (26)
- The "Pulse Sequence" Rules Everything (3)

Seventh Inning Stretch

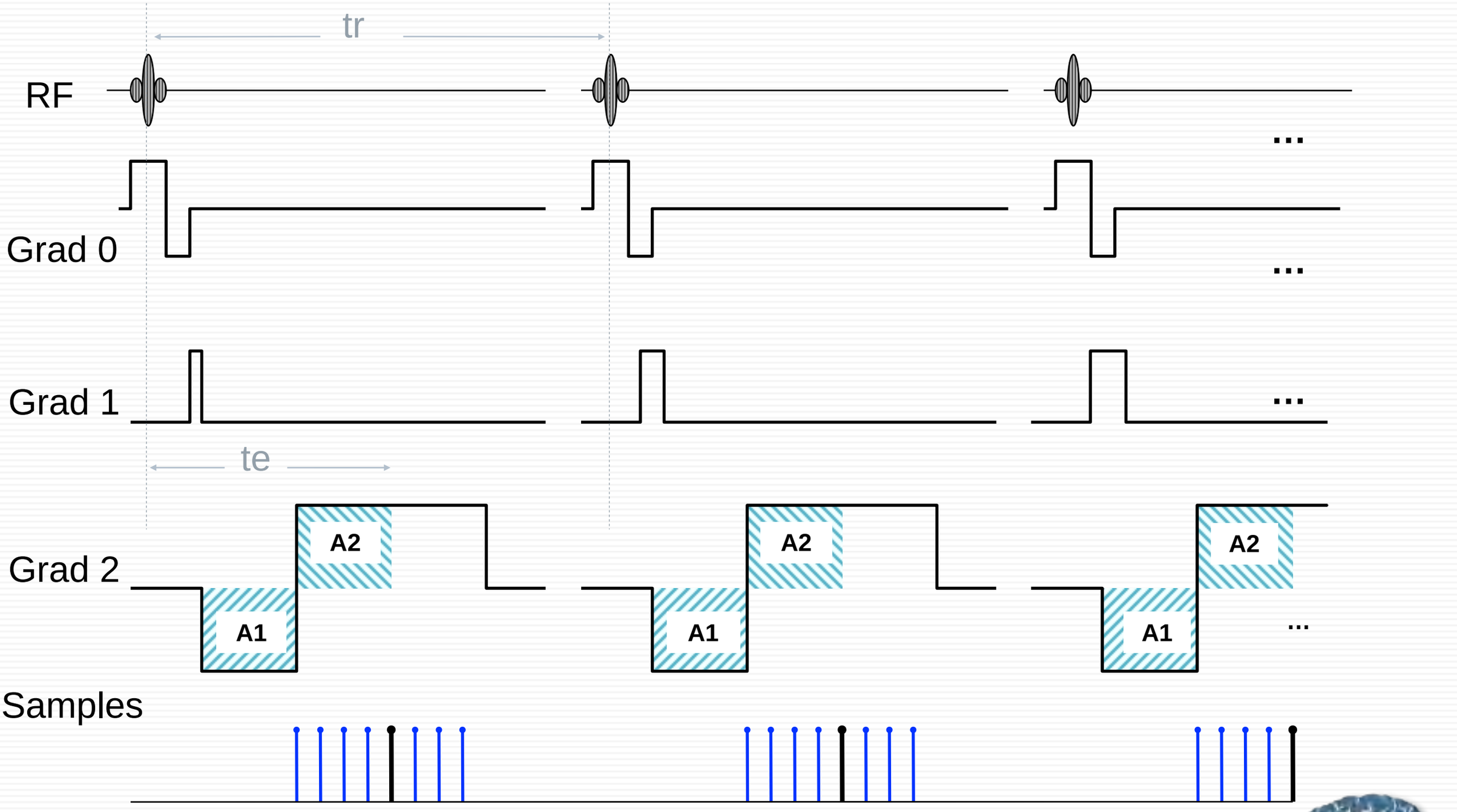
- Fast Imaging (14)
- Functional MRI (18)
- Diffusion and Summary (9)

- Image Quality and Artifacts (48)

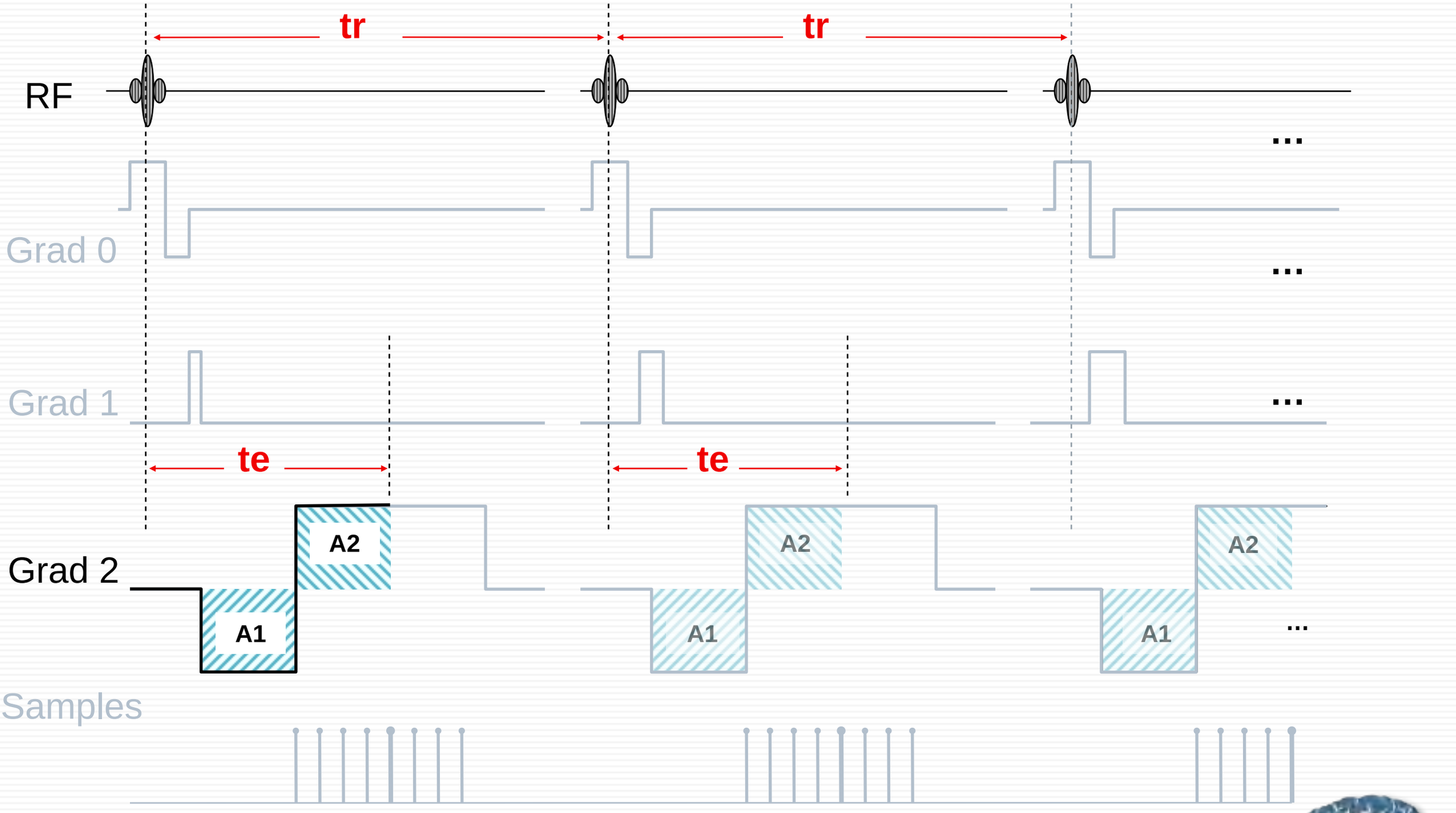
A Pulse Sequence Controls

- Slice Location
- Slice Orientation
- Slice Thickness
- Number of Slices
- Resolution
  (FOV and Matrix)
- Contrast
  TR, TE, TI, Flip Angle, Diffusion, etc…
- Artifact Correction
  Saturation Pulses, Flow Comp,
  Fat Suppression, etc…

Conventional Spatial Encoding



Contrast Encoding



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Reduced Flip Angle Imaging

Outline

- Determinants of Imaging Time
- TR, Saturation and Image Quality
- Reduced Flip Angle Techniques
  - FLASH (=SPGR)
  - FISP (=GRASS)
- Gradient Echoes
- Applications of Shallow Flip Imaging
- Ultra-Fast Imaging

Determinants of Imaging Time

Scan Time =

Repetition Time (TR)
x Number of Phase Encodes
x NEX (Averages)
x Number of 3D Steps

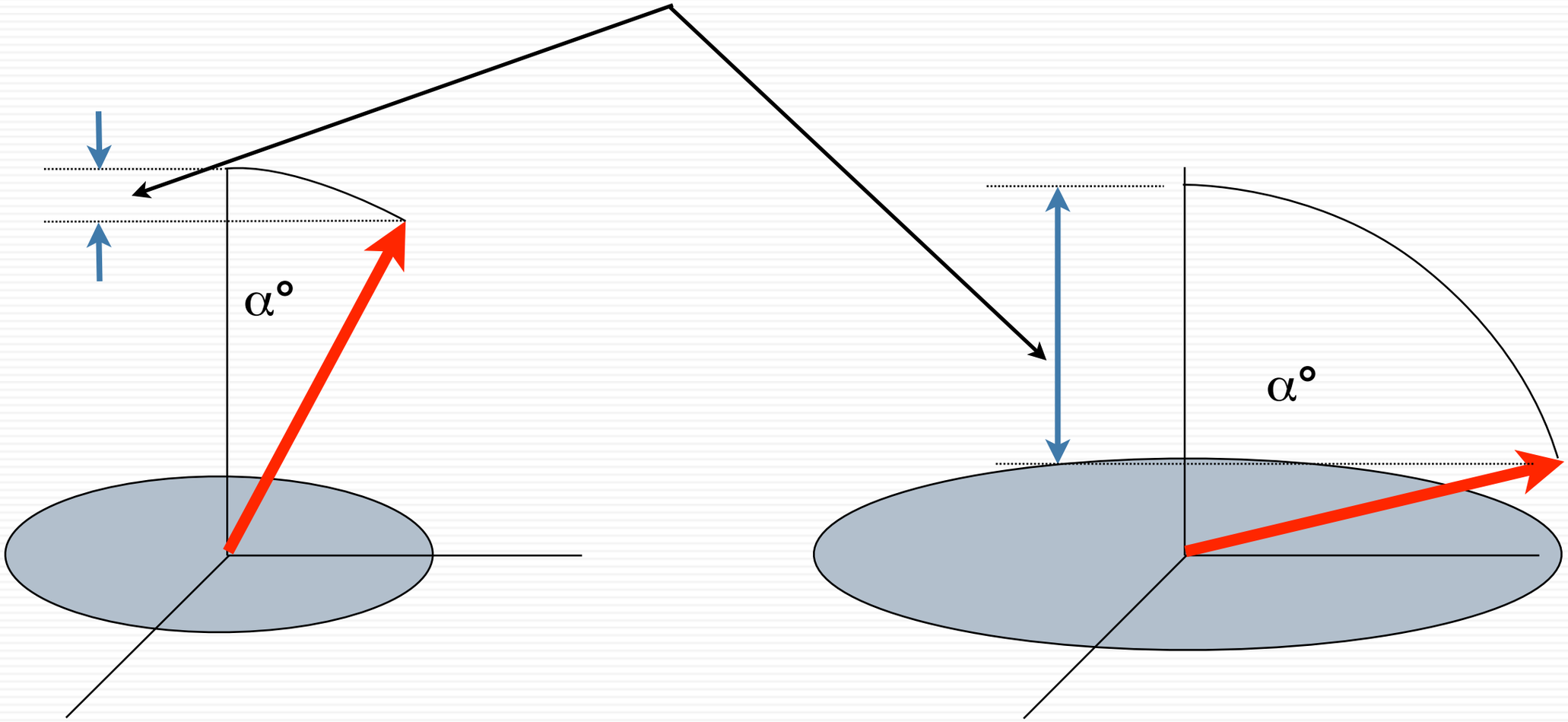
TR and Image Quality

Reduced TR Yields:

- Decreased Scan Time
- Increased T1 Contrast
- Reduced (Useable) T2 Contrast
- Reduced Signal to Noise Ratio
- Increased Power Deposition
- Reduced Slice Coverage

Signal and Flip Angle

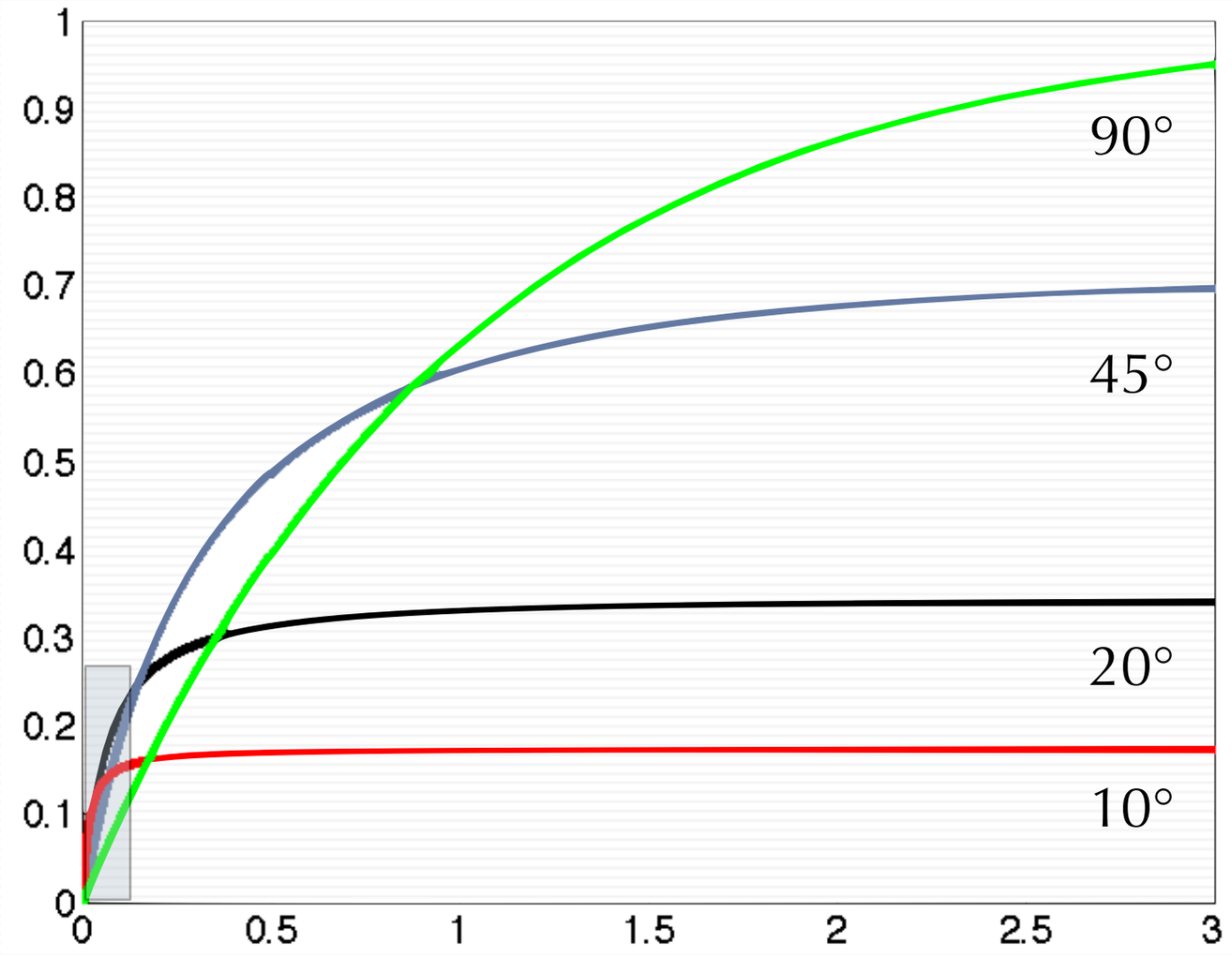
Loss of Longitudinal Magnetization



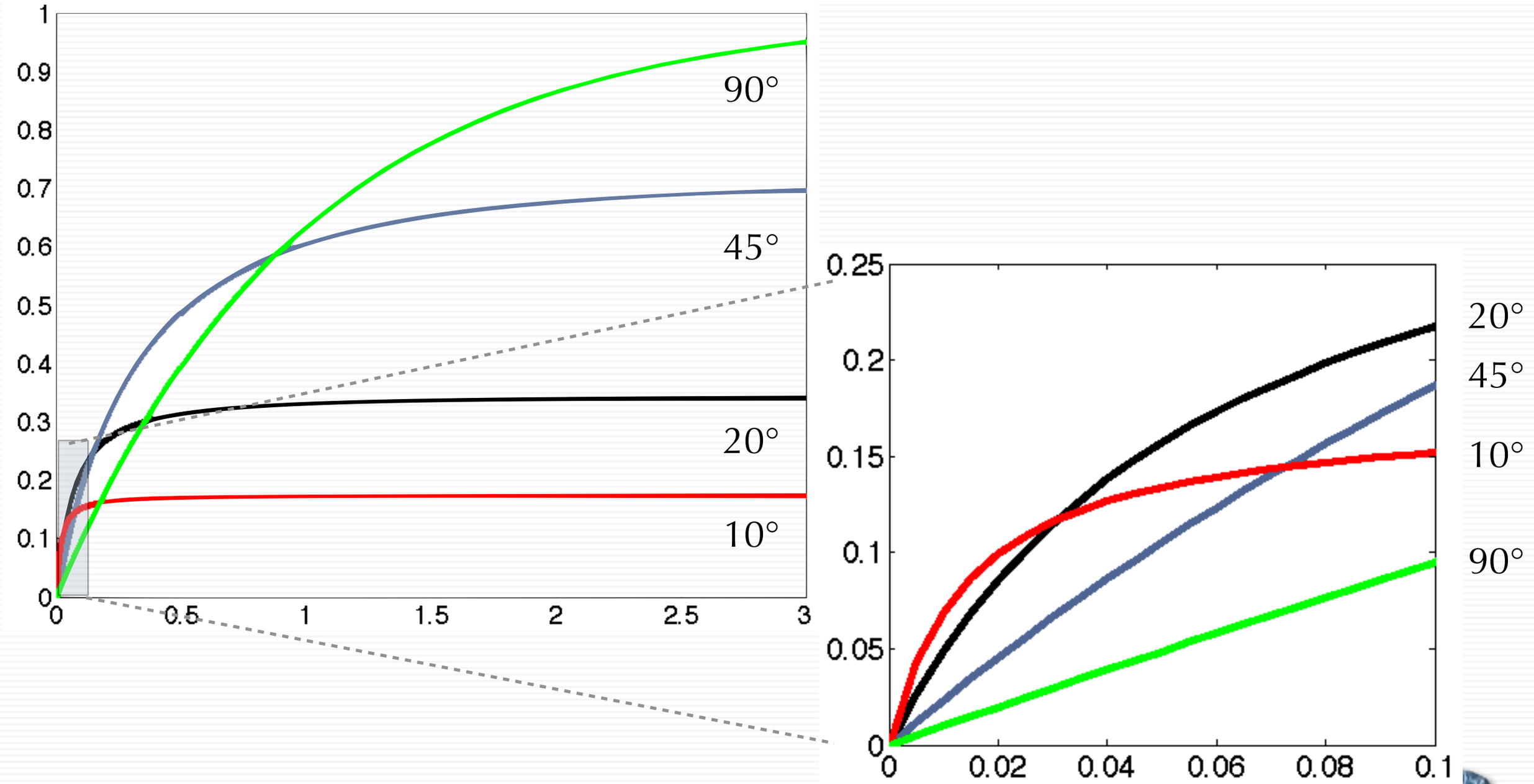
Small Flip Angle

Large Flip Angle

Flip Angle and TR/T1



Flip Angle and TR/T1

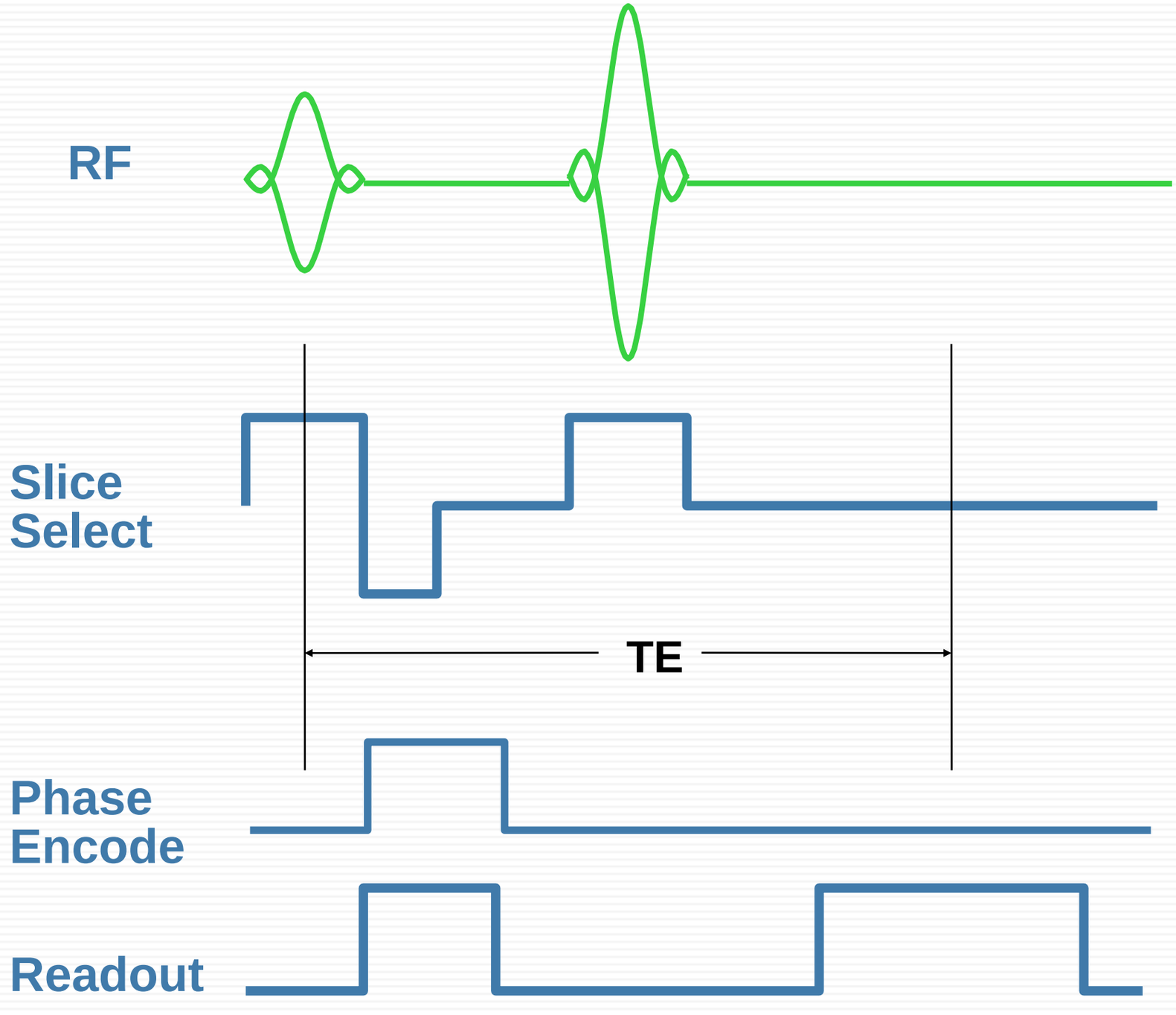


Contrast and Flip Angle

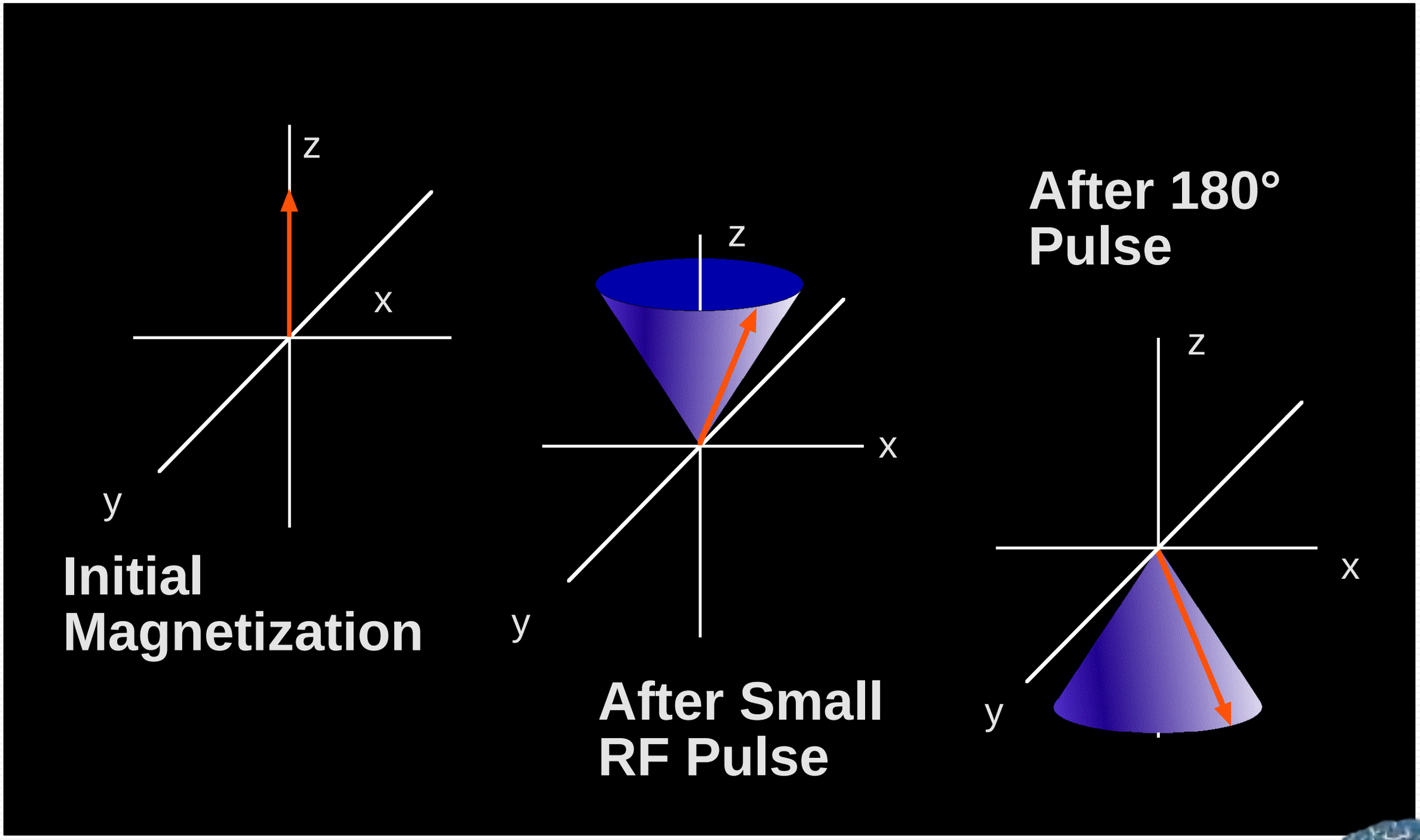
Large Flip Angles	Short	Long
Long	Proton Density	T2* Weighted
Short	T1 Weighted	

Small Flip Angles	Short	Long
Long	Proton Density	T2* Weighted
Short	Proton Density	T2* Weighted

Spin Echo Sequence



A 180° Pulse is not used in FLASH imaging



T2: Transverse Magnetization Decay from Spin-Spin Interactions

T2*: Transverse Magnetization Decay from Local Magnetic Field Variations

Magnetic Susceptibility

The Extent to Which a Substance Becomes "MAGNETIZED" when Placed Within a Magnetic Field

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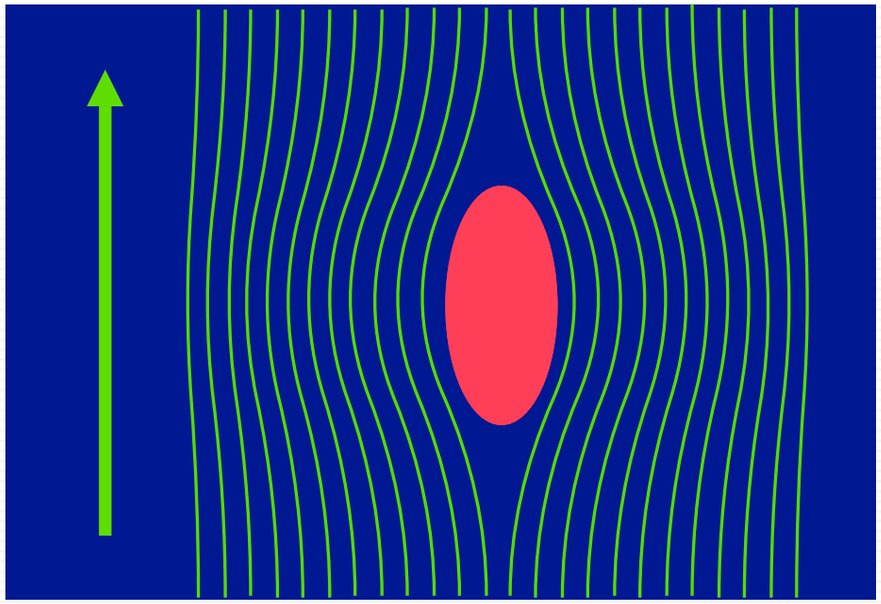
The Extent to Which a Substance Becomes "MAGNETIZED" when Placed Within a Magnetic Field

Applied Magnetic Field

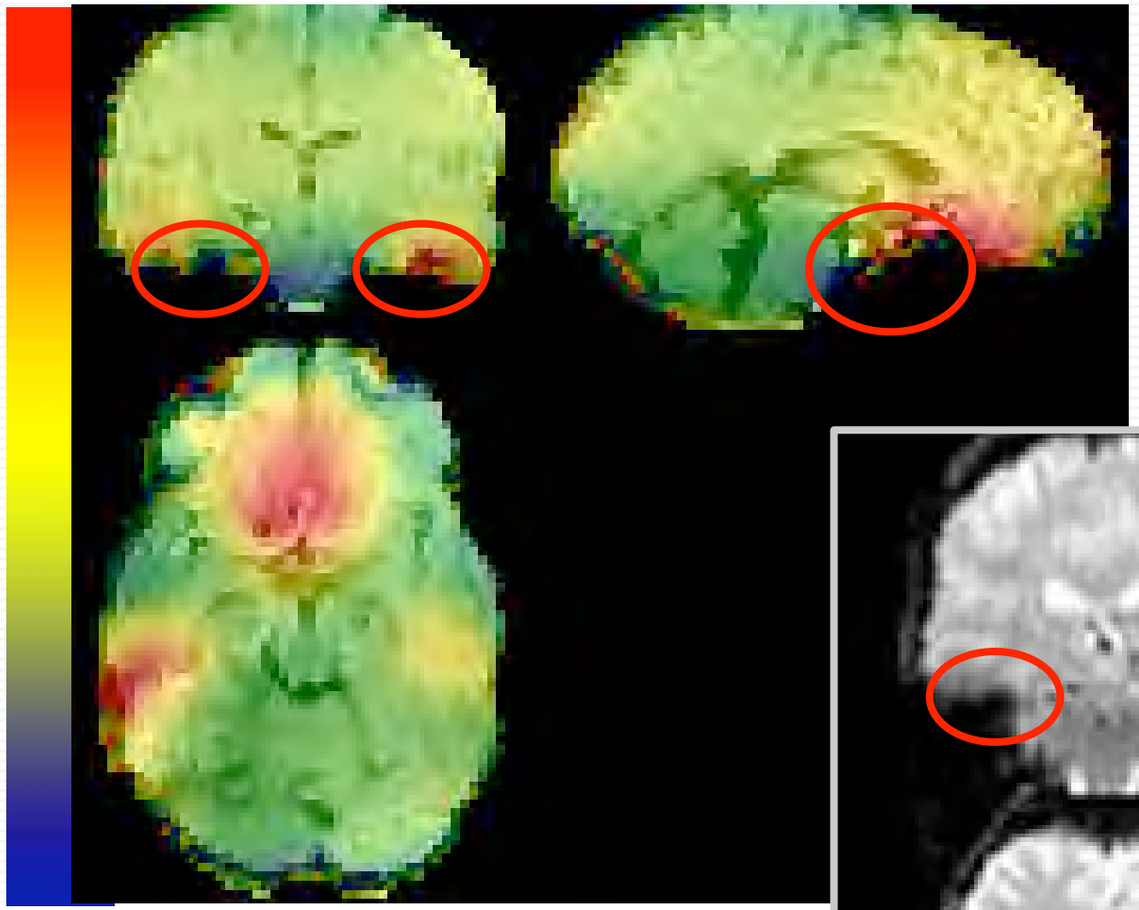


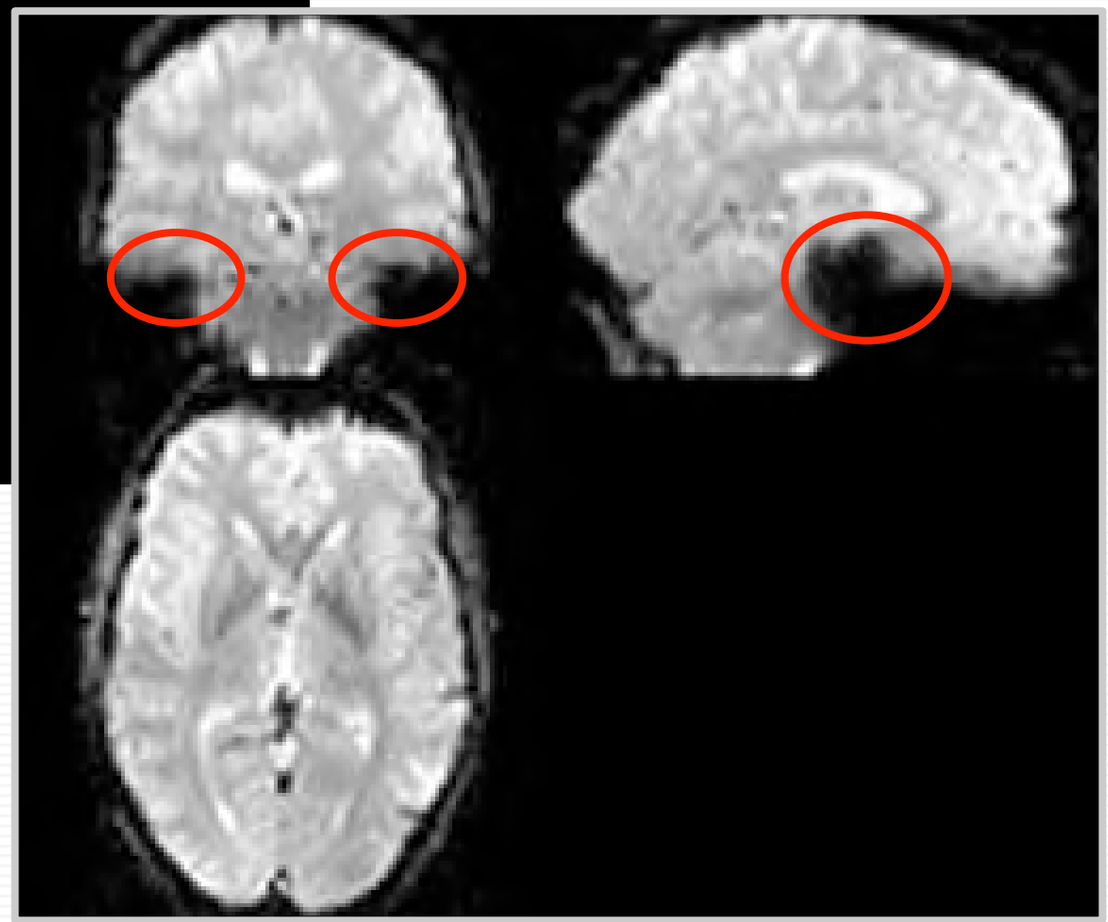
Objects with Susceptibility Different than Air Distort the Magnetic Field

Local field Variations Result in Signal Loss

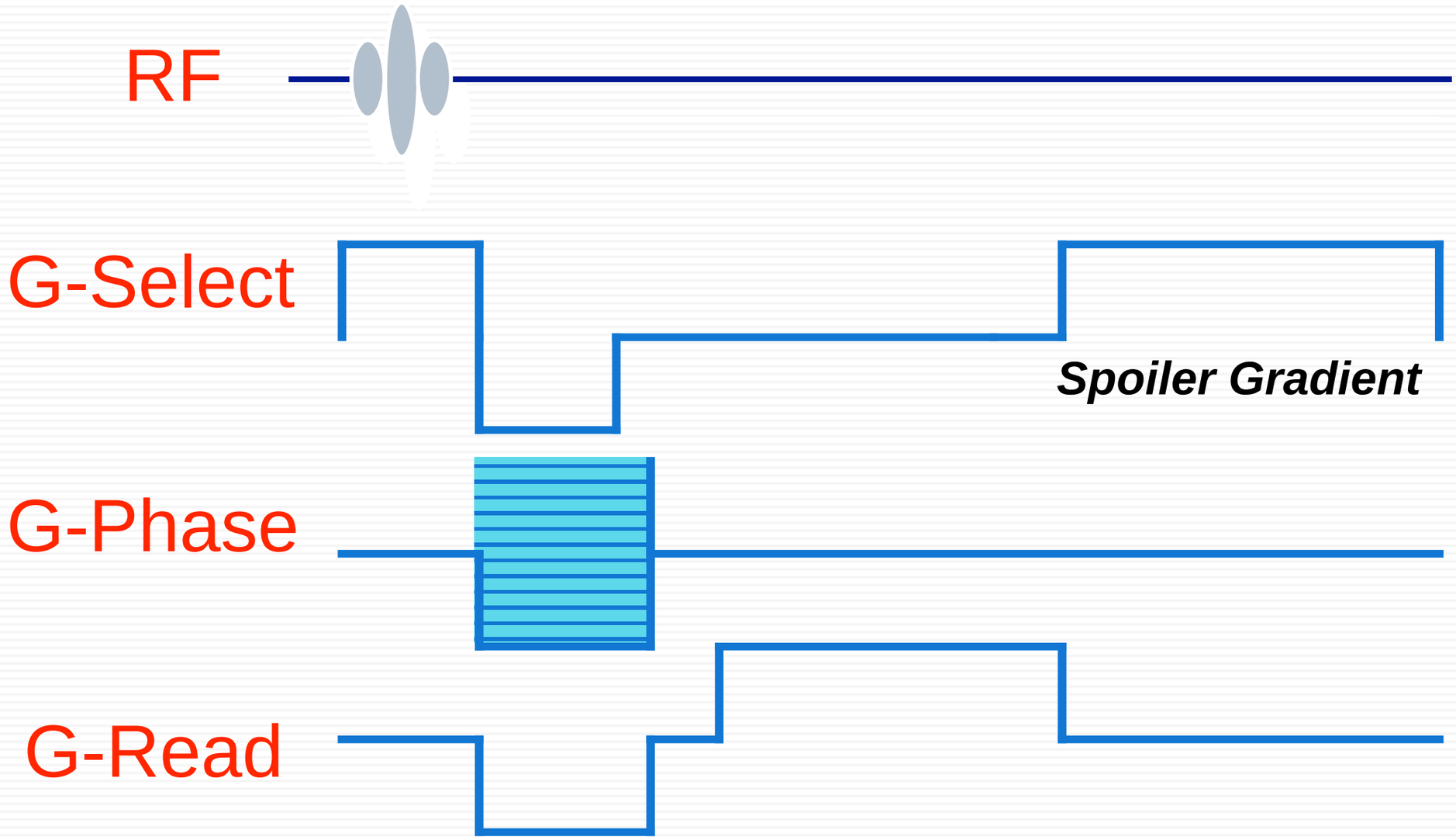
+3 ppm

-3 ppm

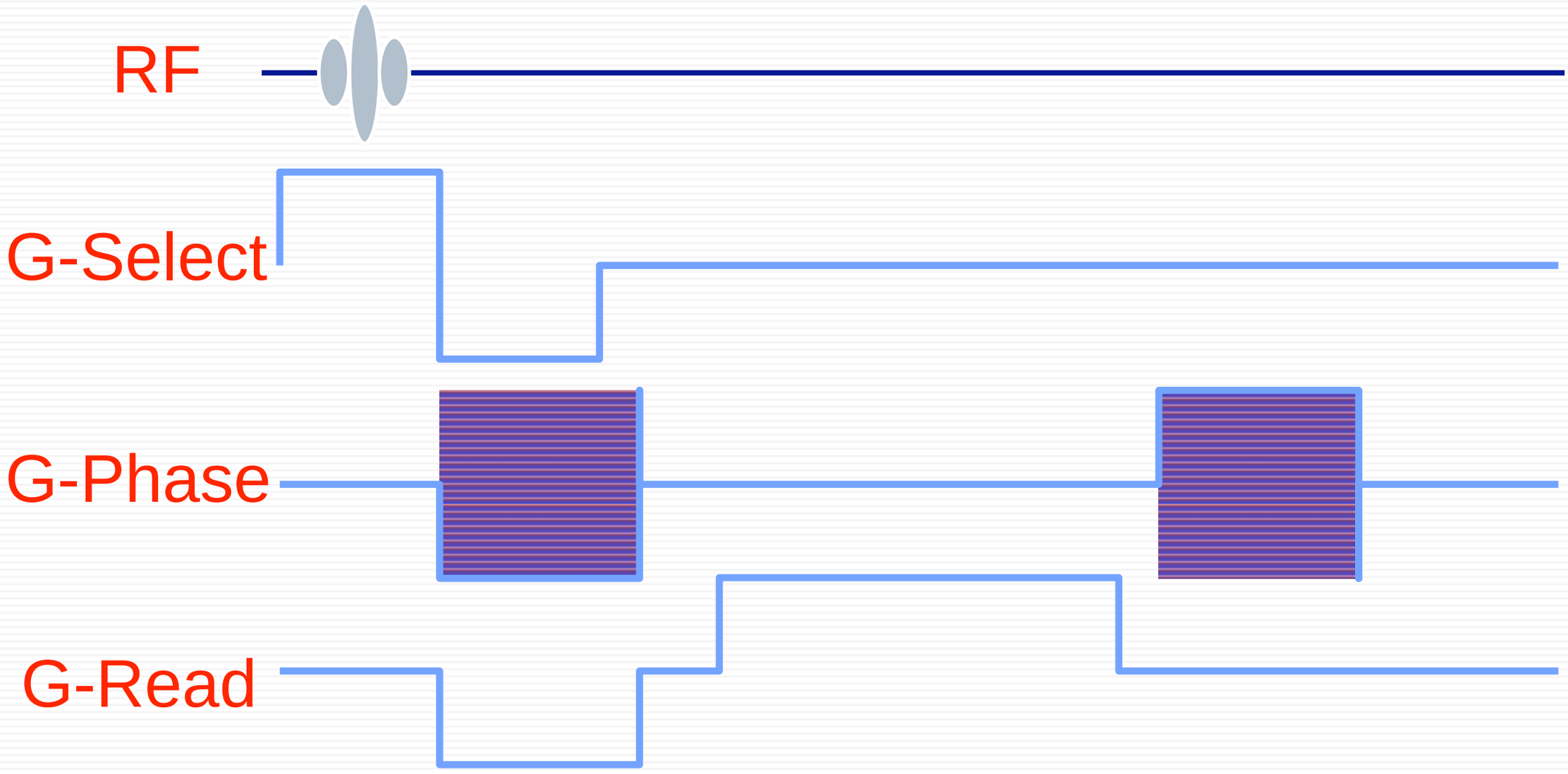




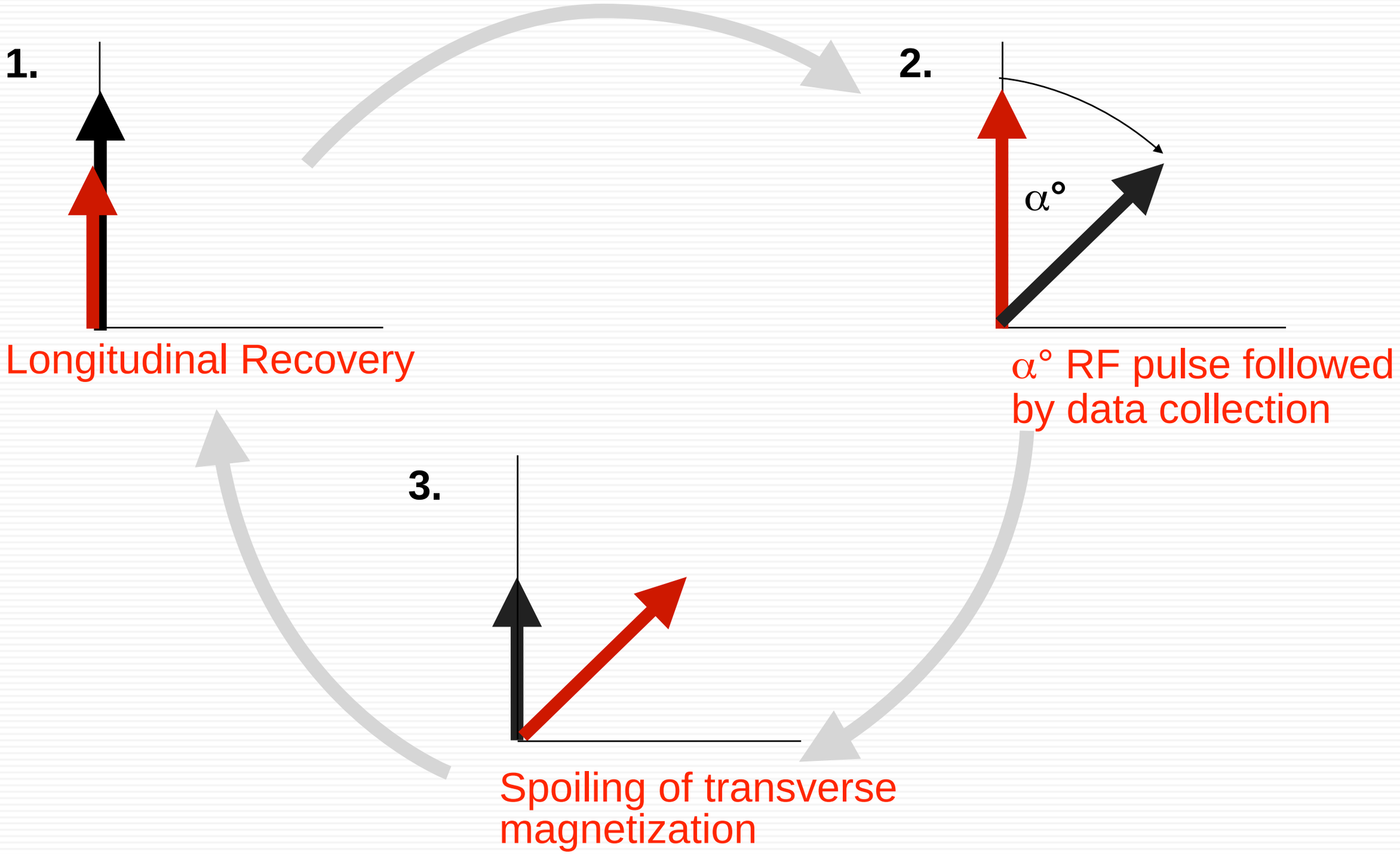
FLASH Timing Diagram



FISP (GRASS) Timing Diagram

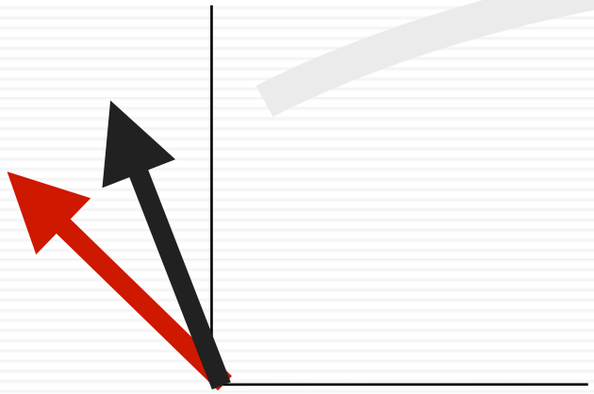


the FLASH Magnetization Cycle



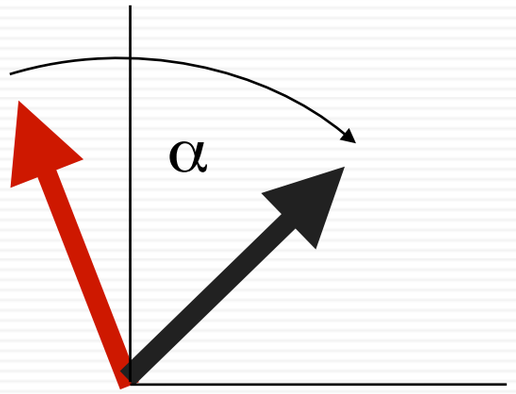
The SSFP Magnetization Cycle

1.



Longitudinal Recovery
and T2* relaxation

2.

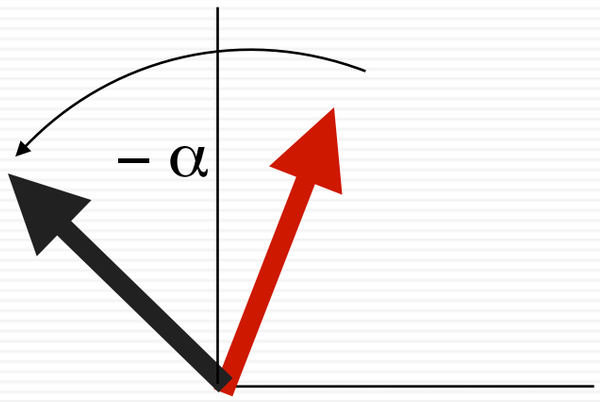


α degree RF pulse
and data collection

3.

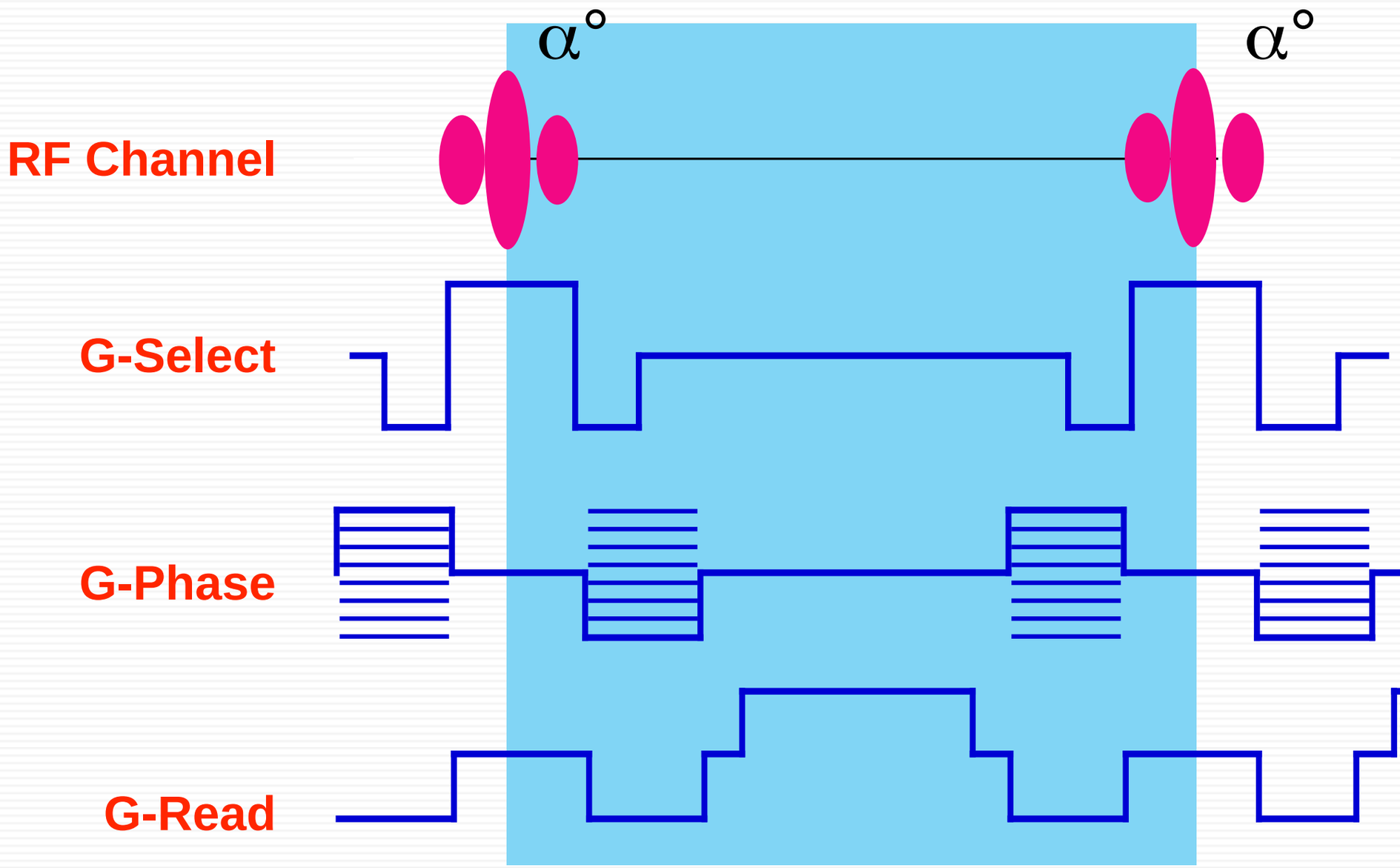
Longitudinal Recovery
and T2* relaxation

4.

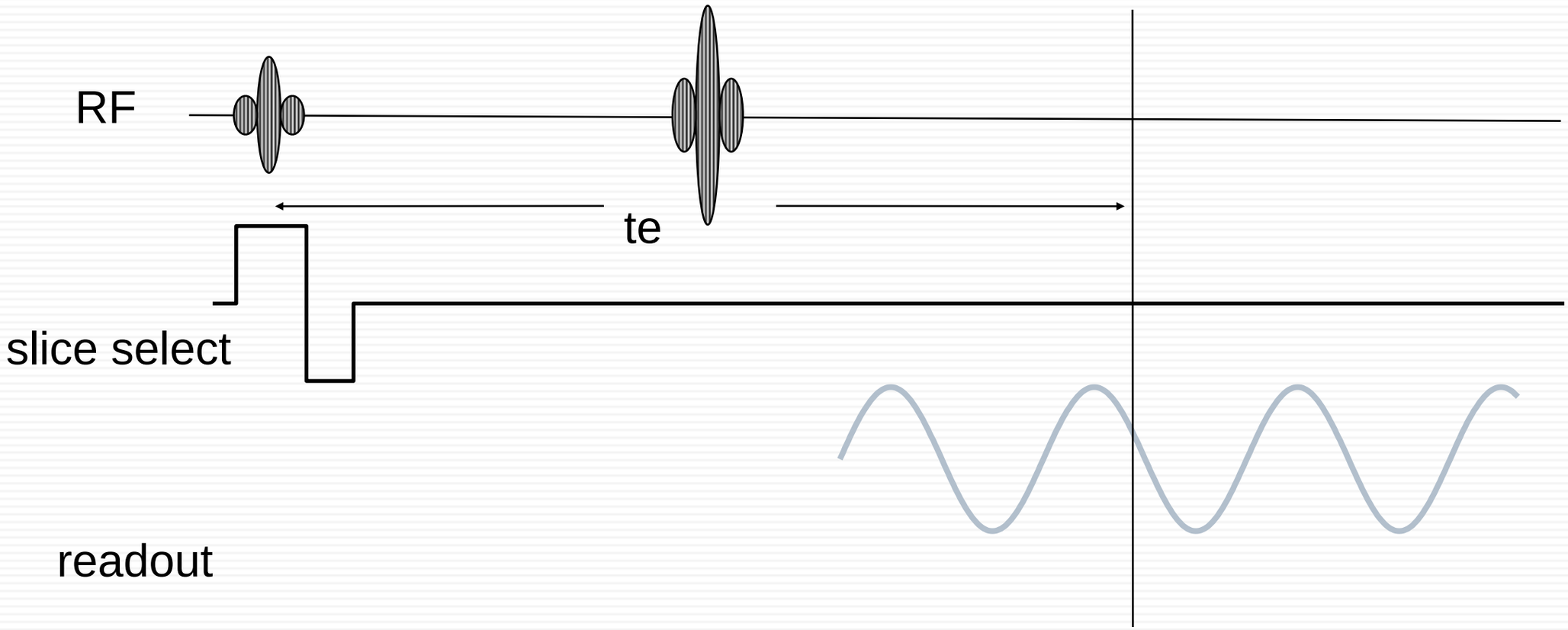


α degree RF pulse
and data collection

SSFP Sequence

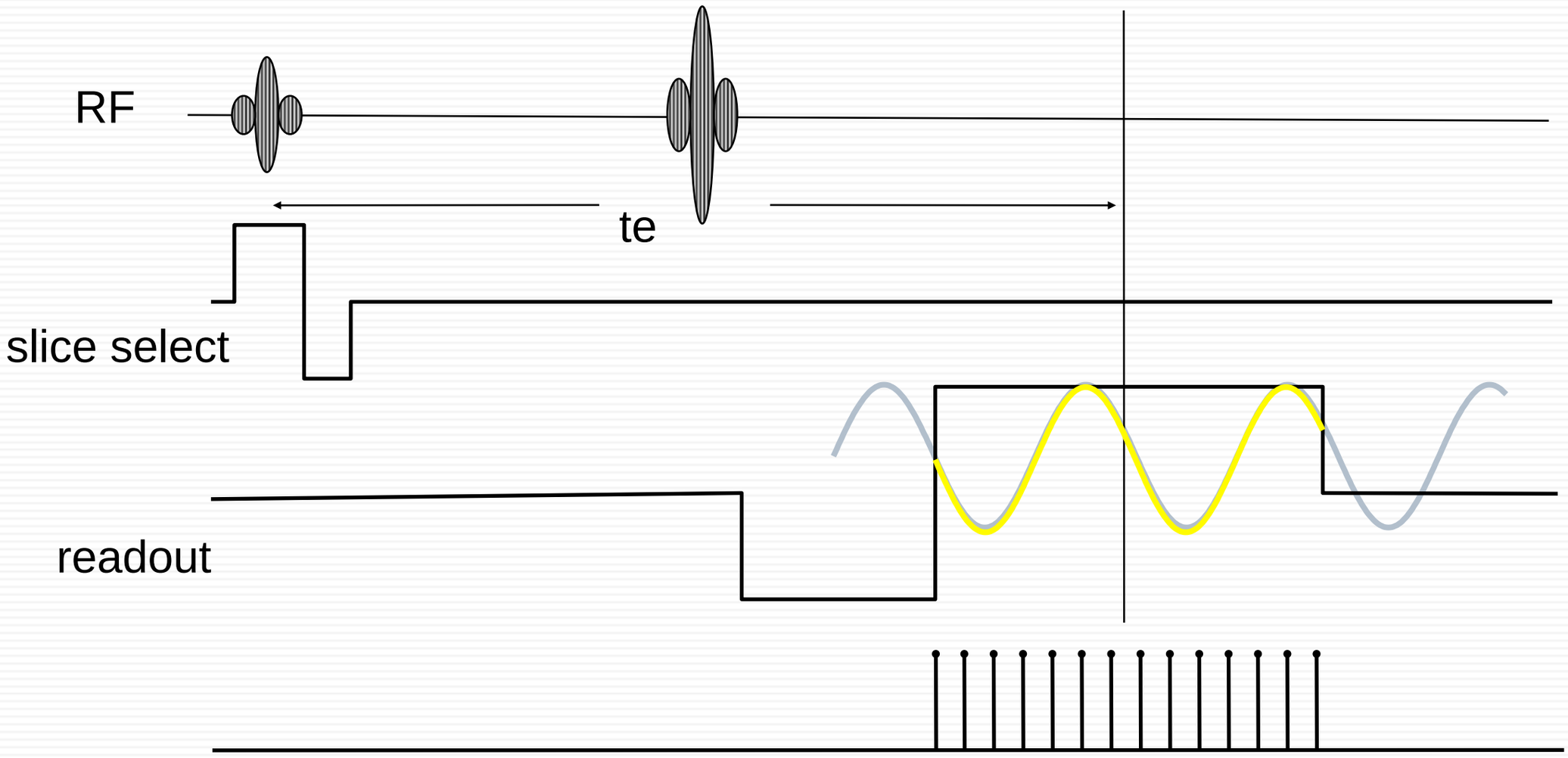


Phase Maps



readout

Phase Maps



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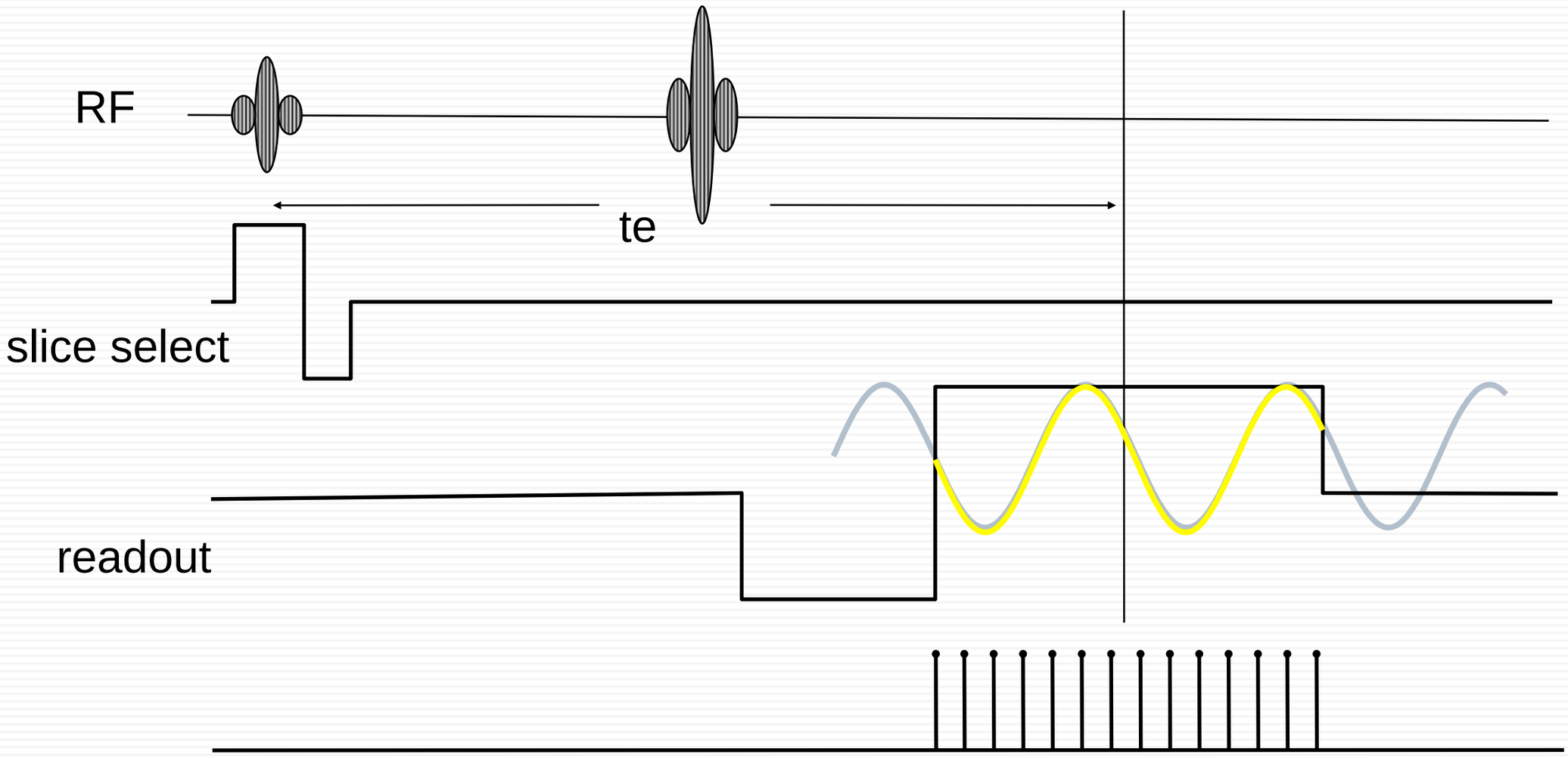

- Time shift in data collection amounts to a phase offset
- Spins precessing at different rates (different magnetic fields) will acquire different phase shifts

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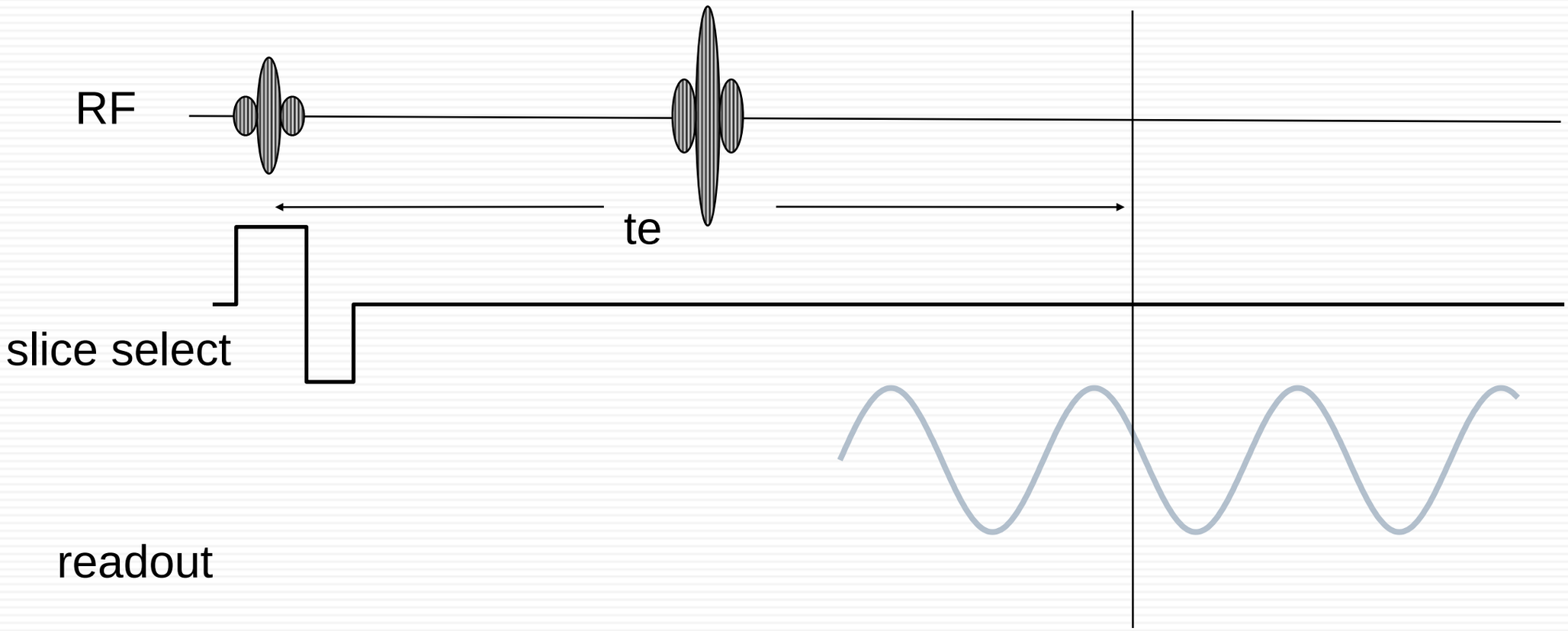

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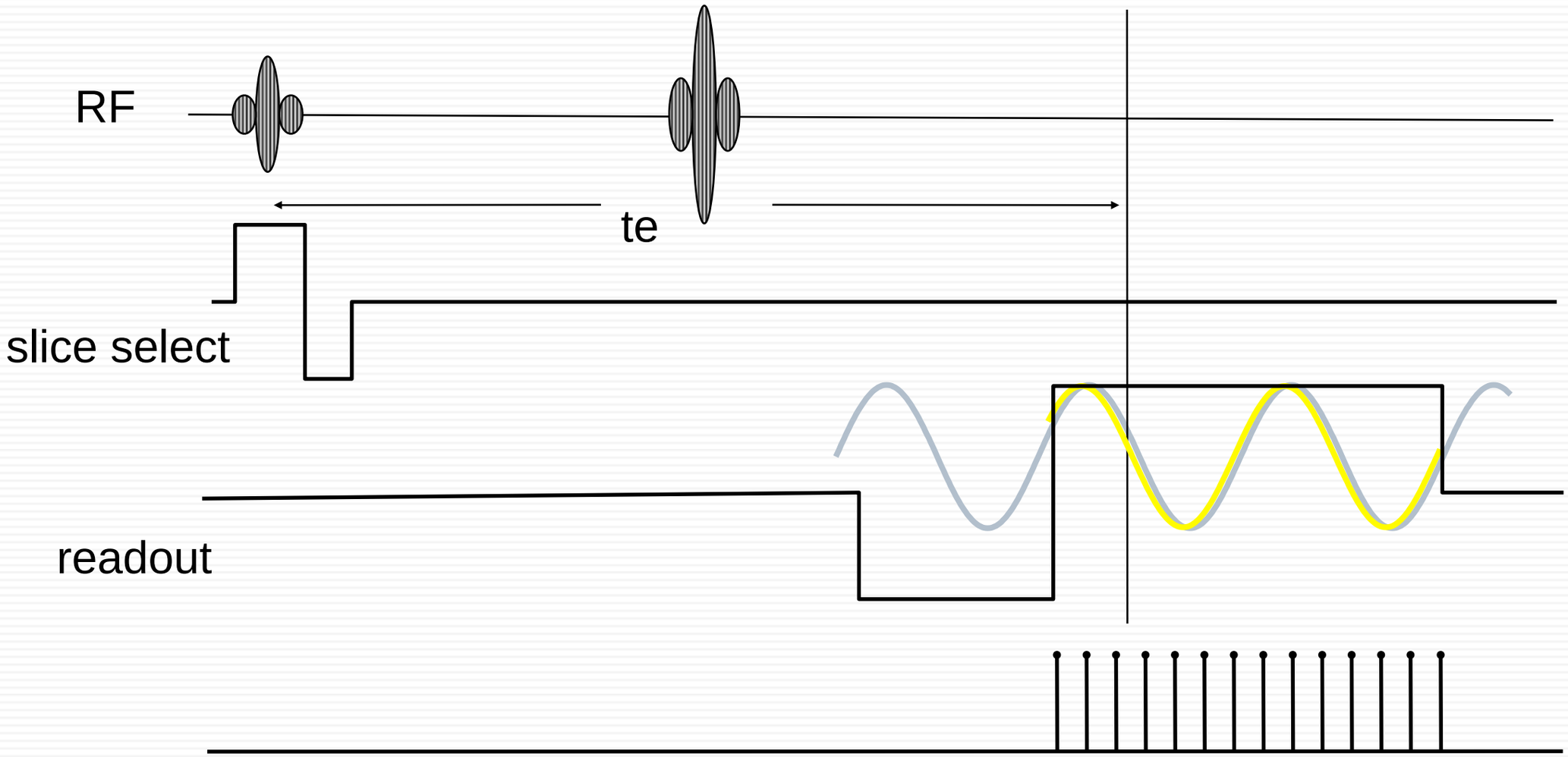

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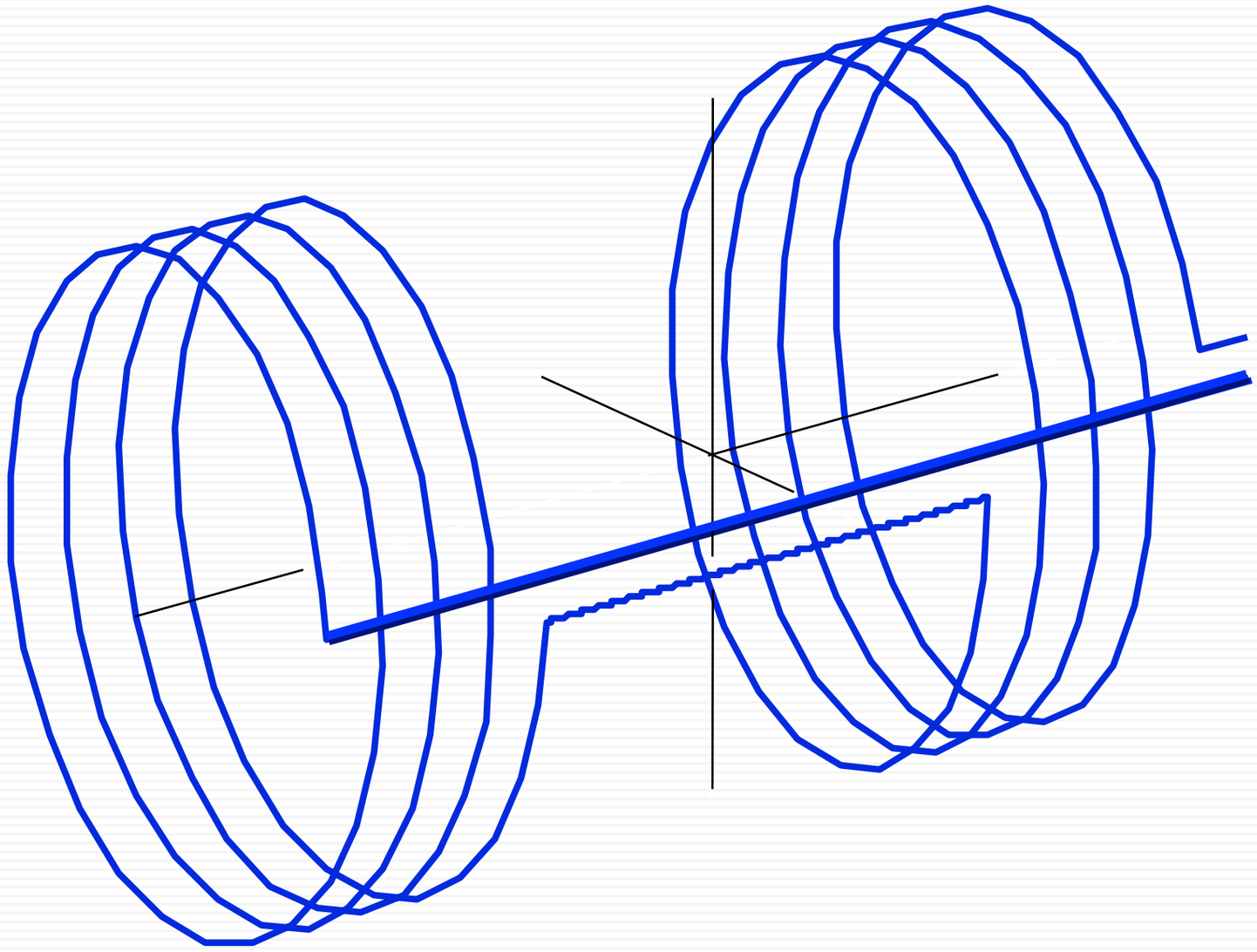
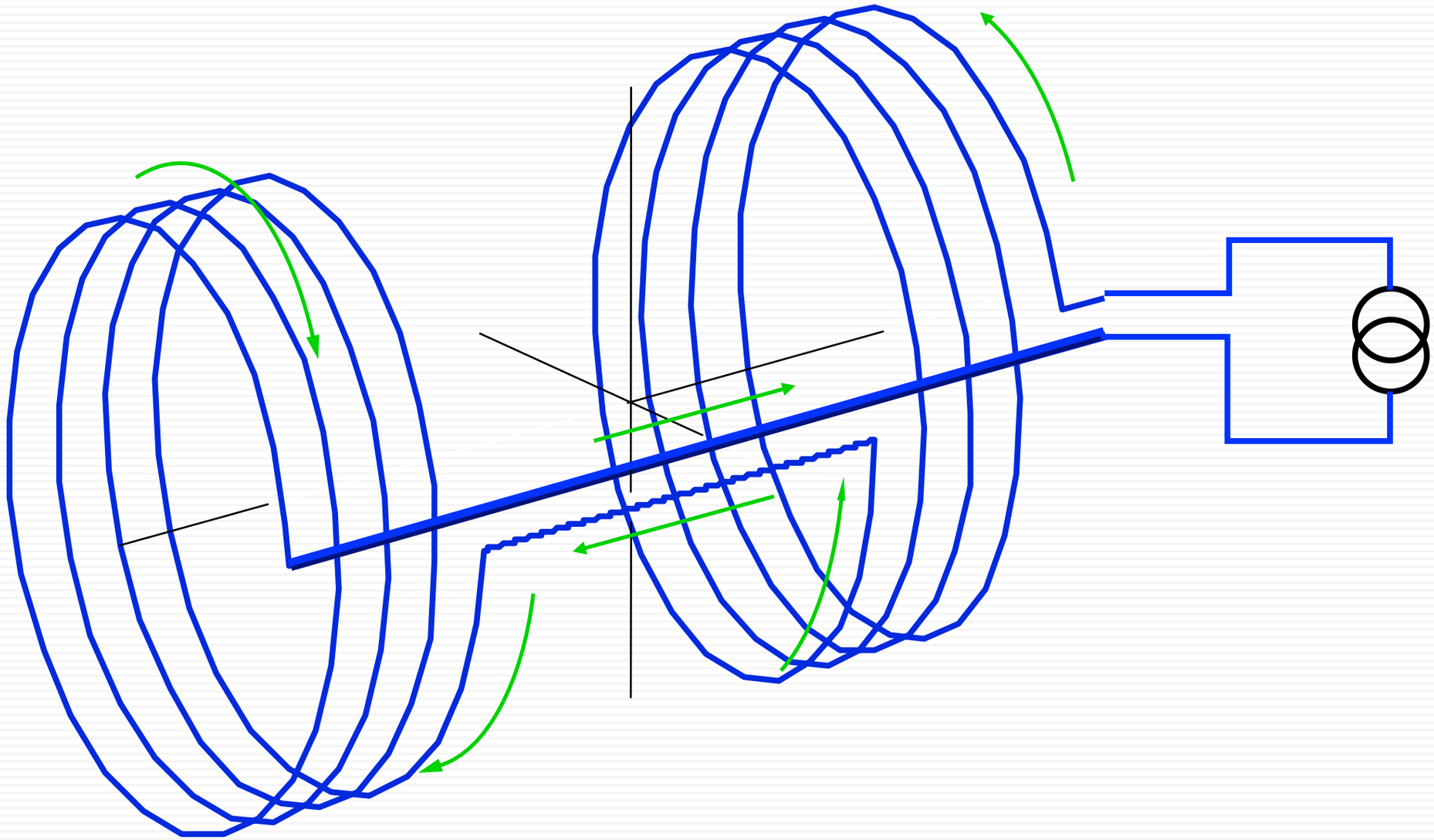
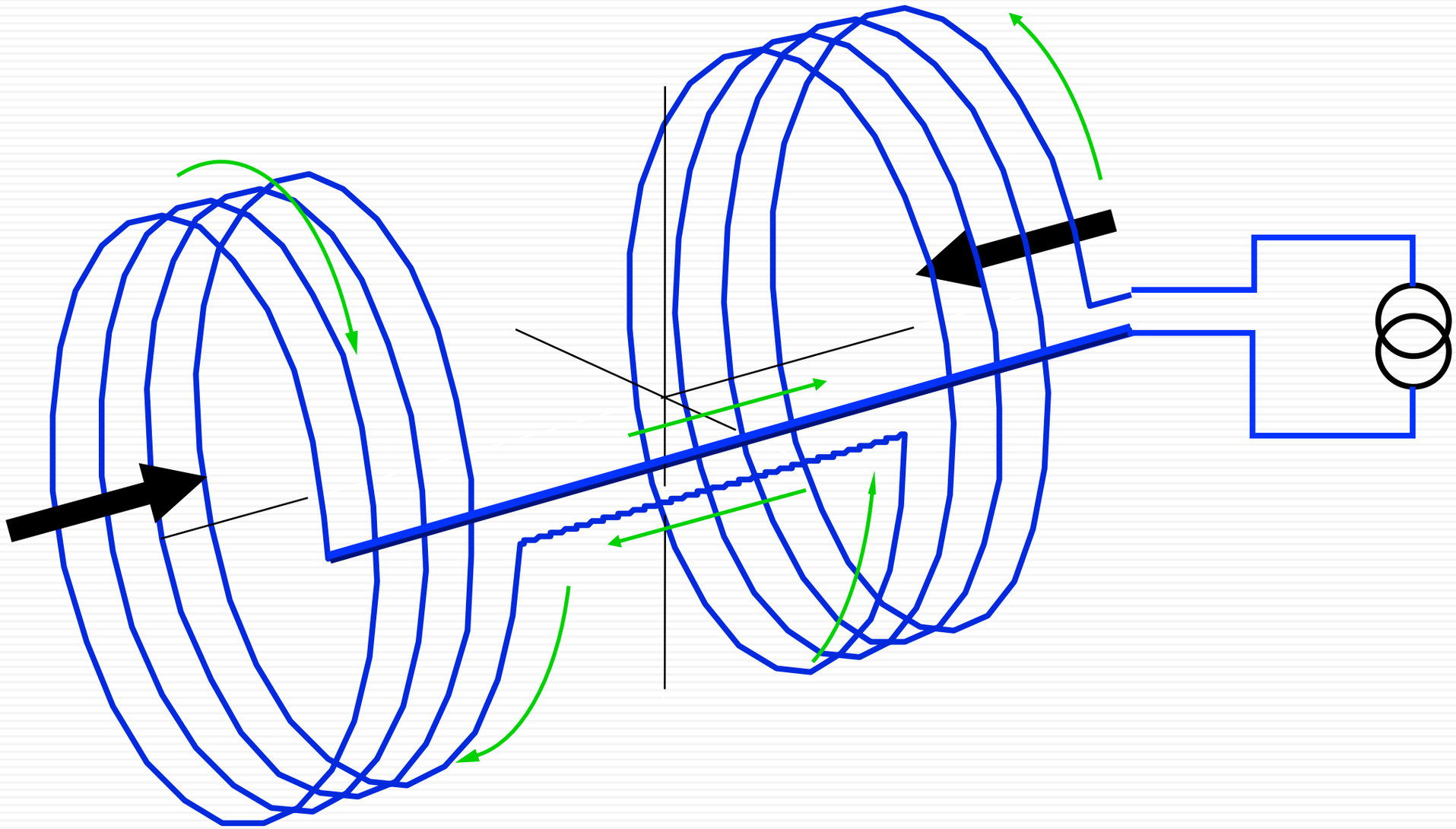
MR Field Gradient Coil



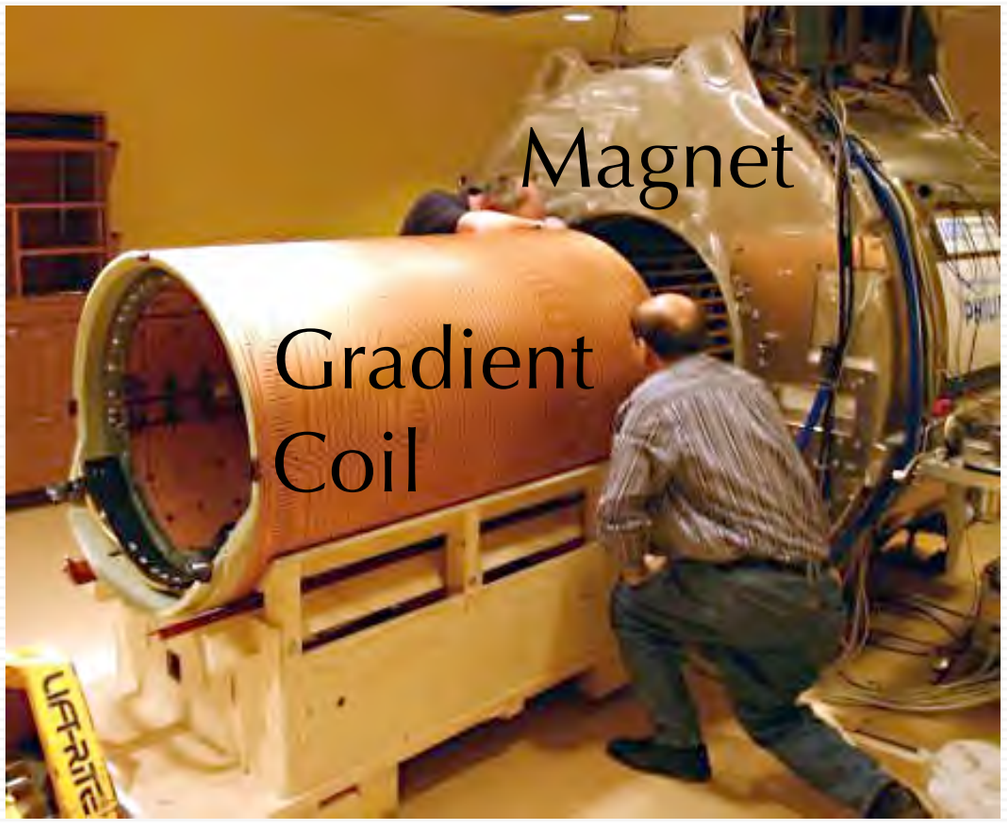
MR Field Gradient Coil

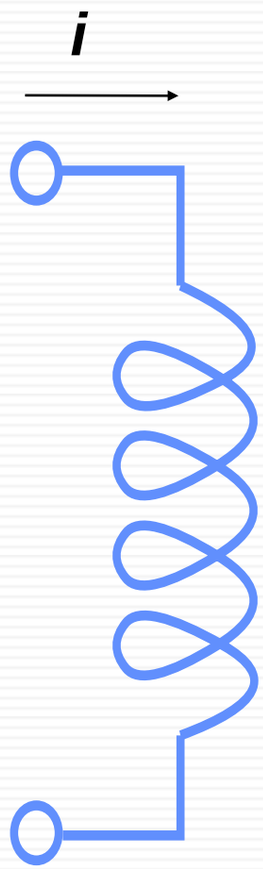


MR Field Gradient Coil



Gradient Coil Characteristics





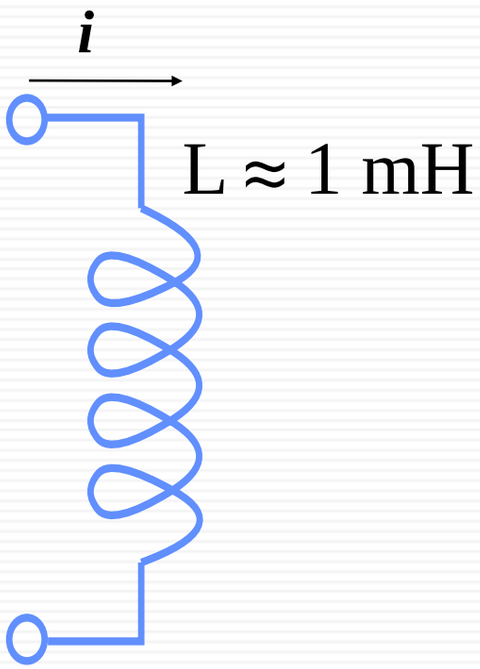
L ≈ 1 mH

Gradient Coil

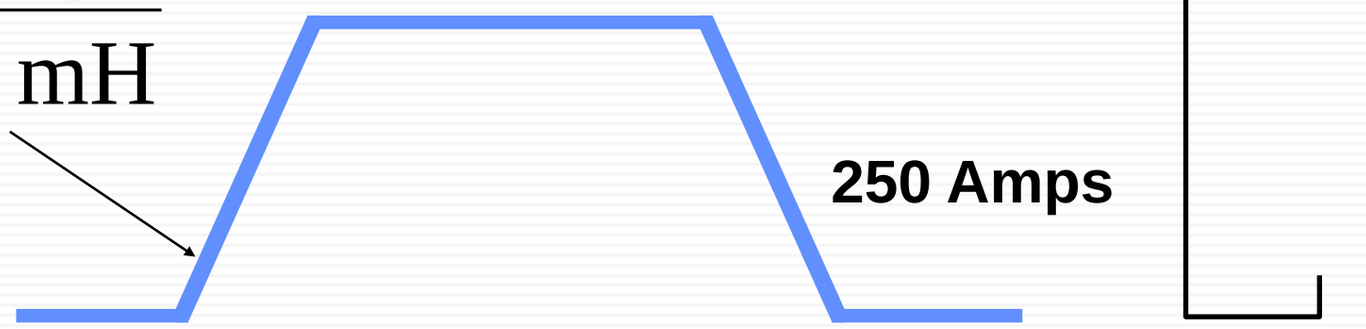
Gradient Strength = k *i*

k ≈ 1 Gauss/cm / 100 Amps

Rise Time, Current and Voltage



$$\frac{di}{dt} = \frac{V_L}{1\ \text{mH}}$$



For: 250 Amps in 100 μsec

V_L = 2500 Volts

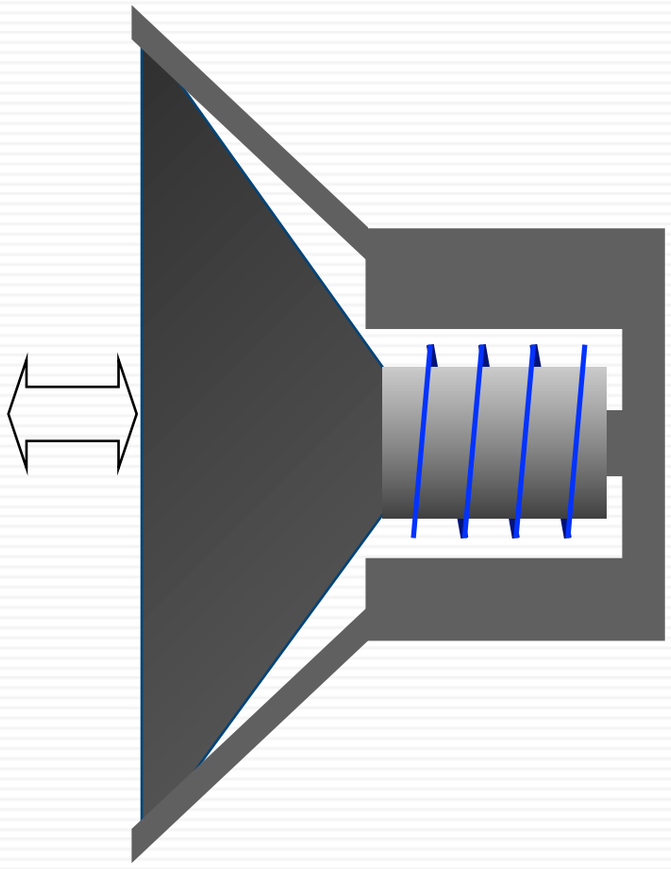
Power = 2500 Volts x 250 Amps

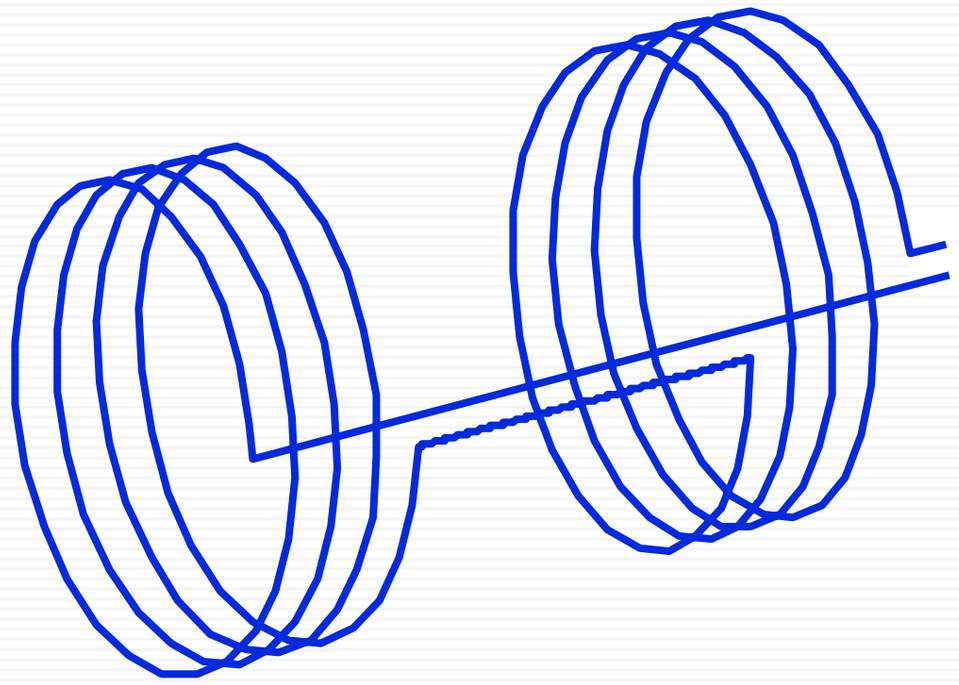
= 6.25×10^5 Watts

Gradient Rise Time strongly affects:

- ✓ Minimum TE
- ✓ Minimum Echo Spacing
- ✓ Echo Planar Bandwidth

Why is MRI So Noisy?





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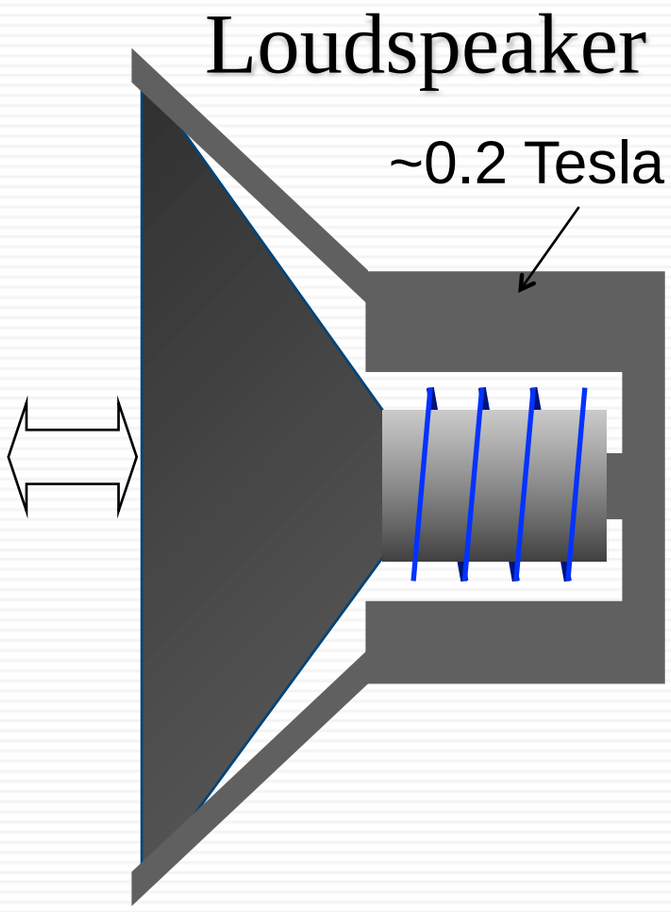
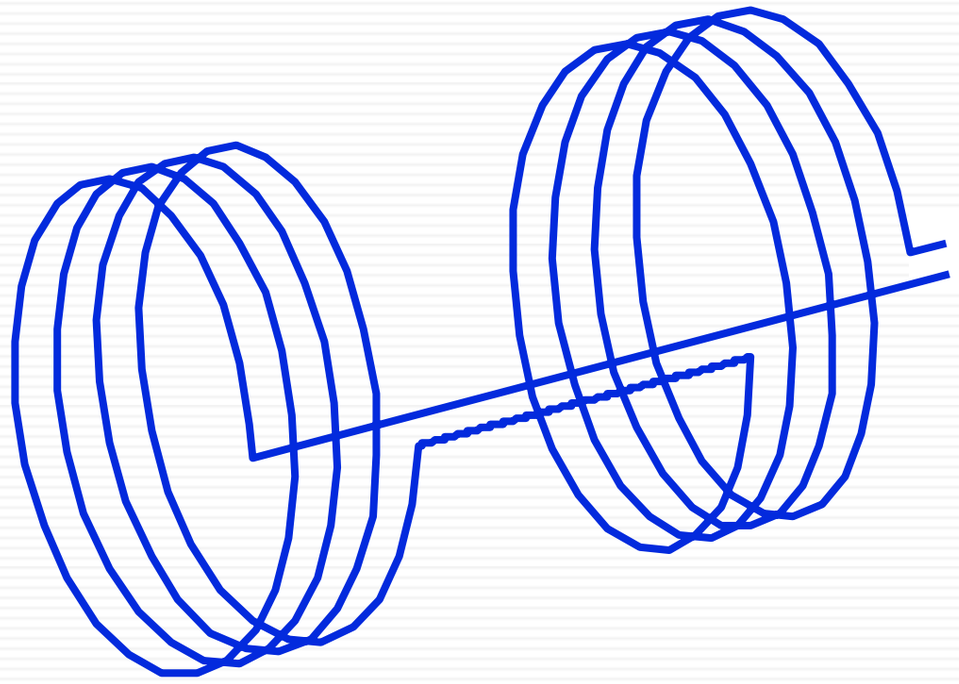
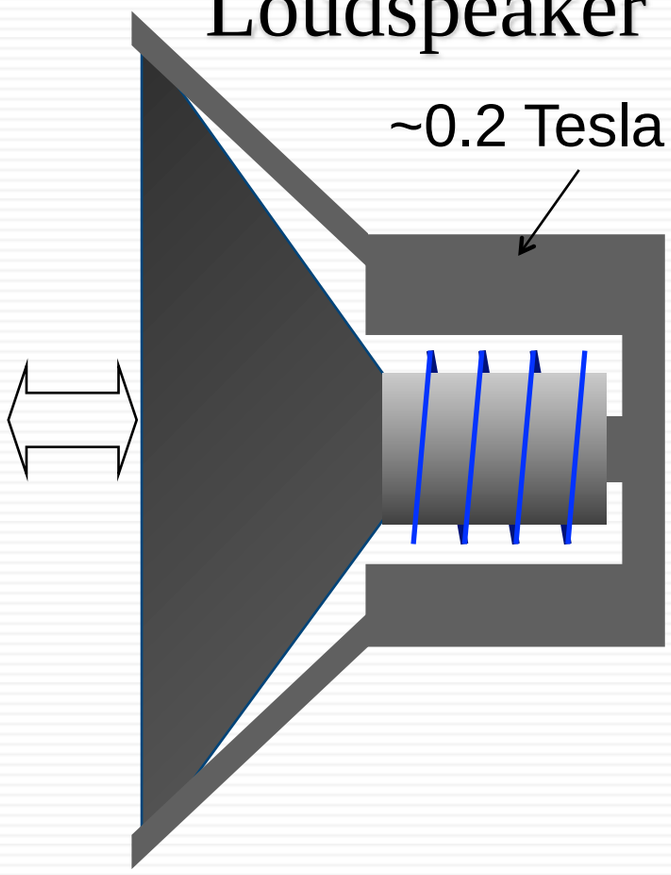


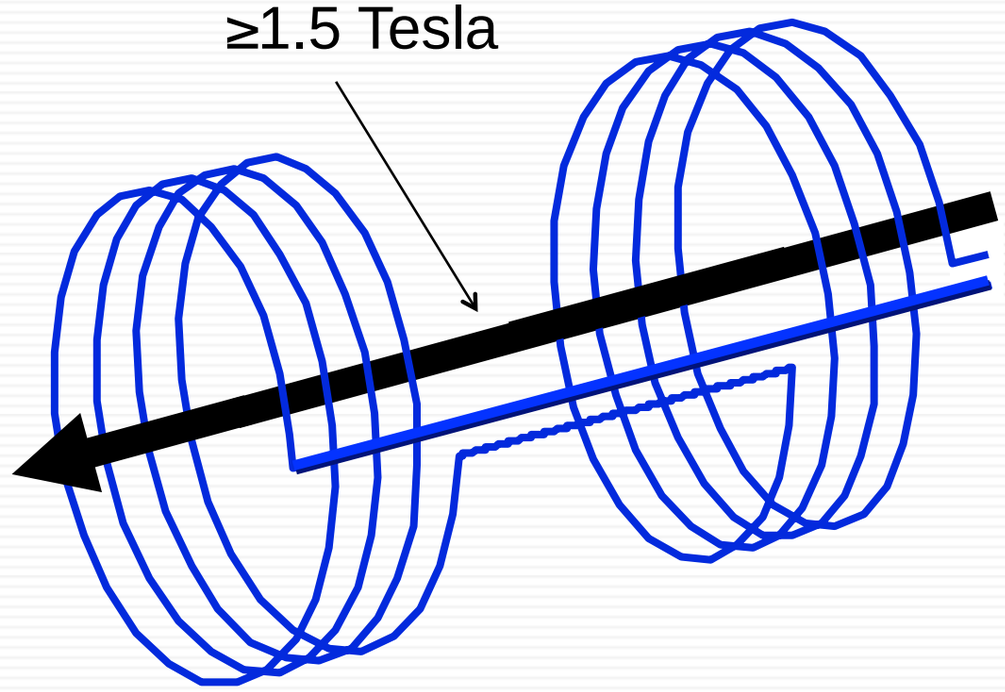


Why is MRI So Noisy?

Loudspeaker



MR Field Gradient



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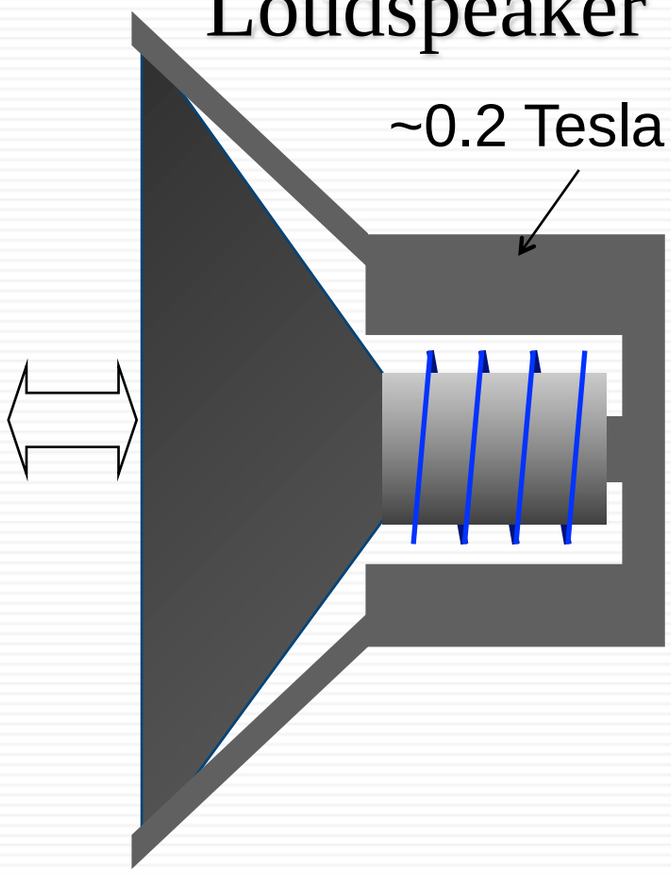
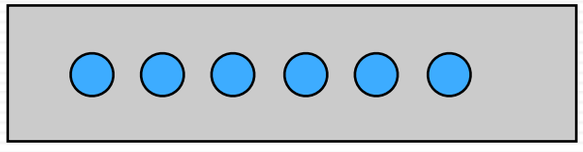
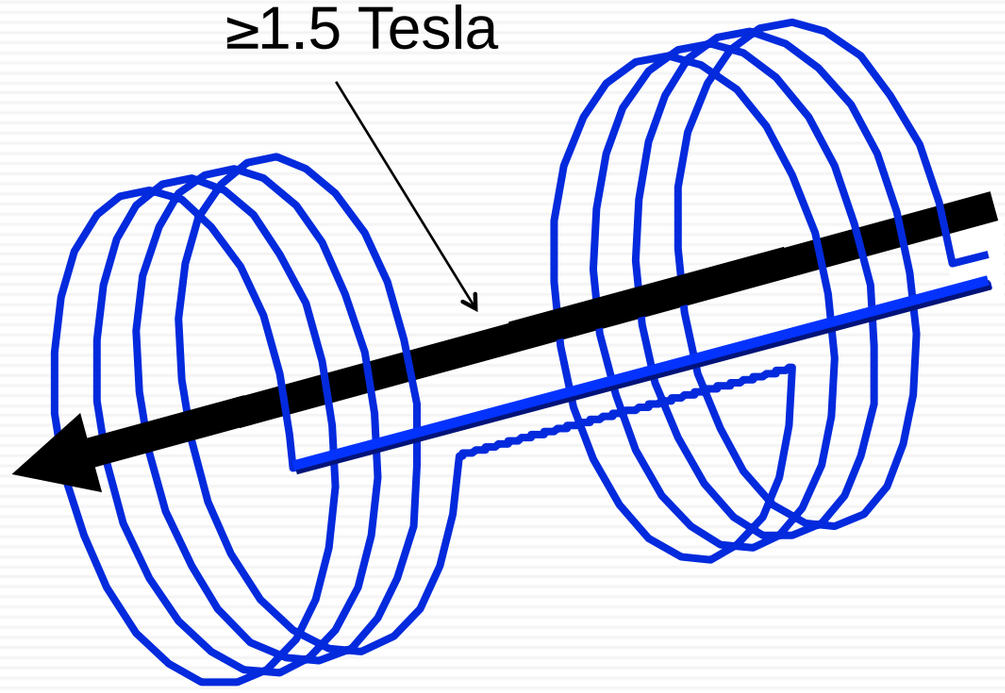




MR Field Gradient



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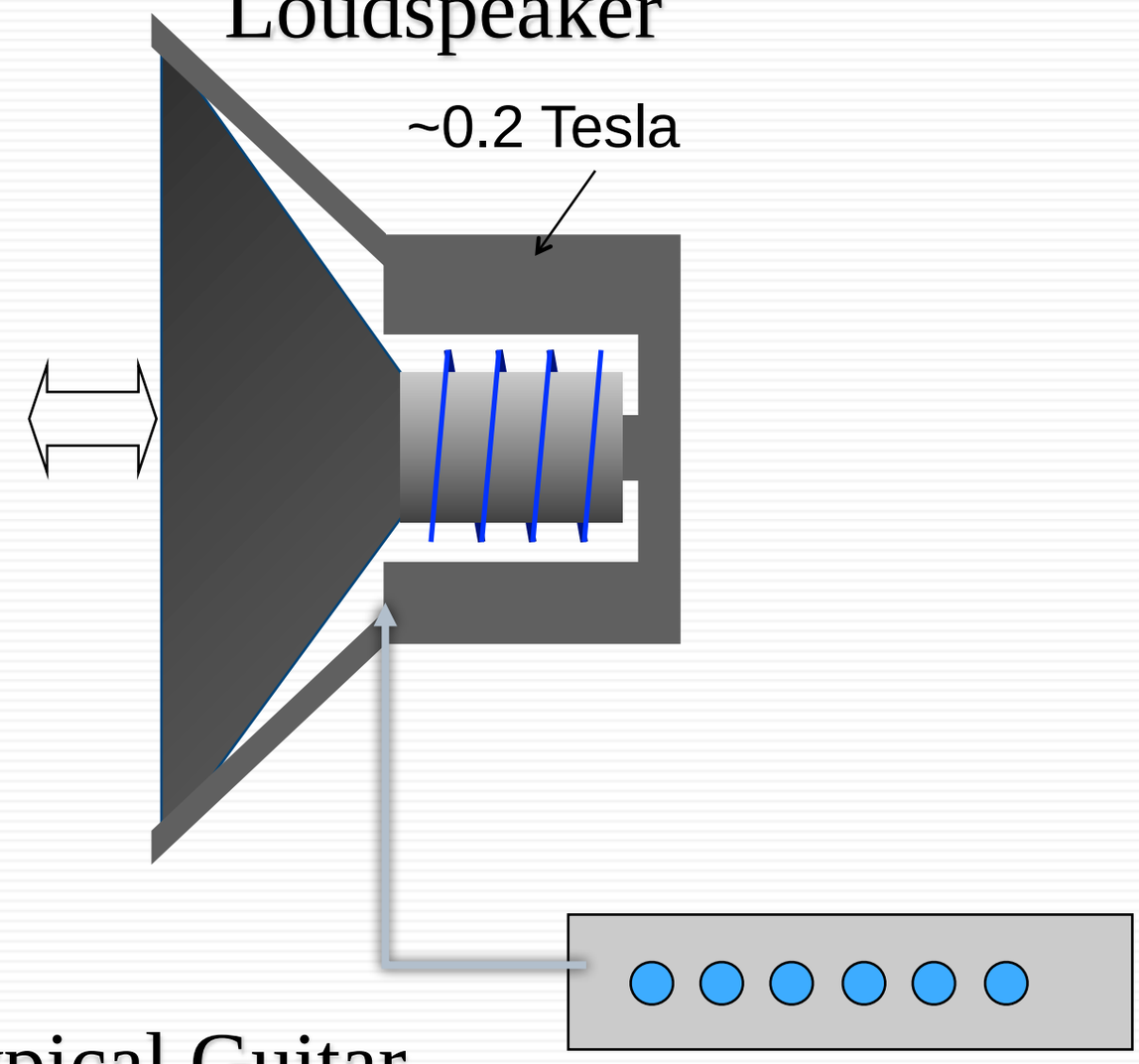

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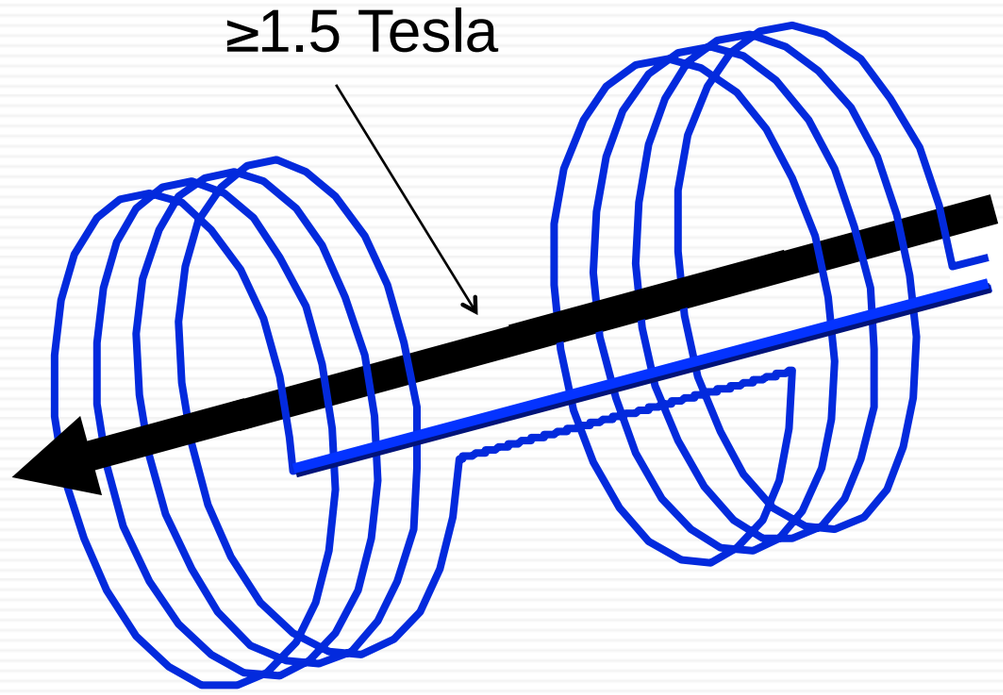
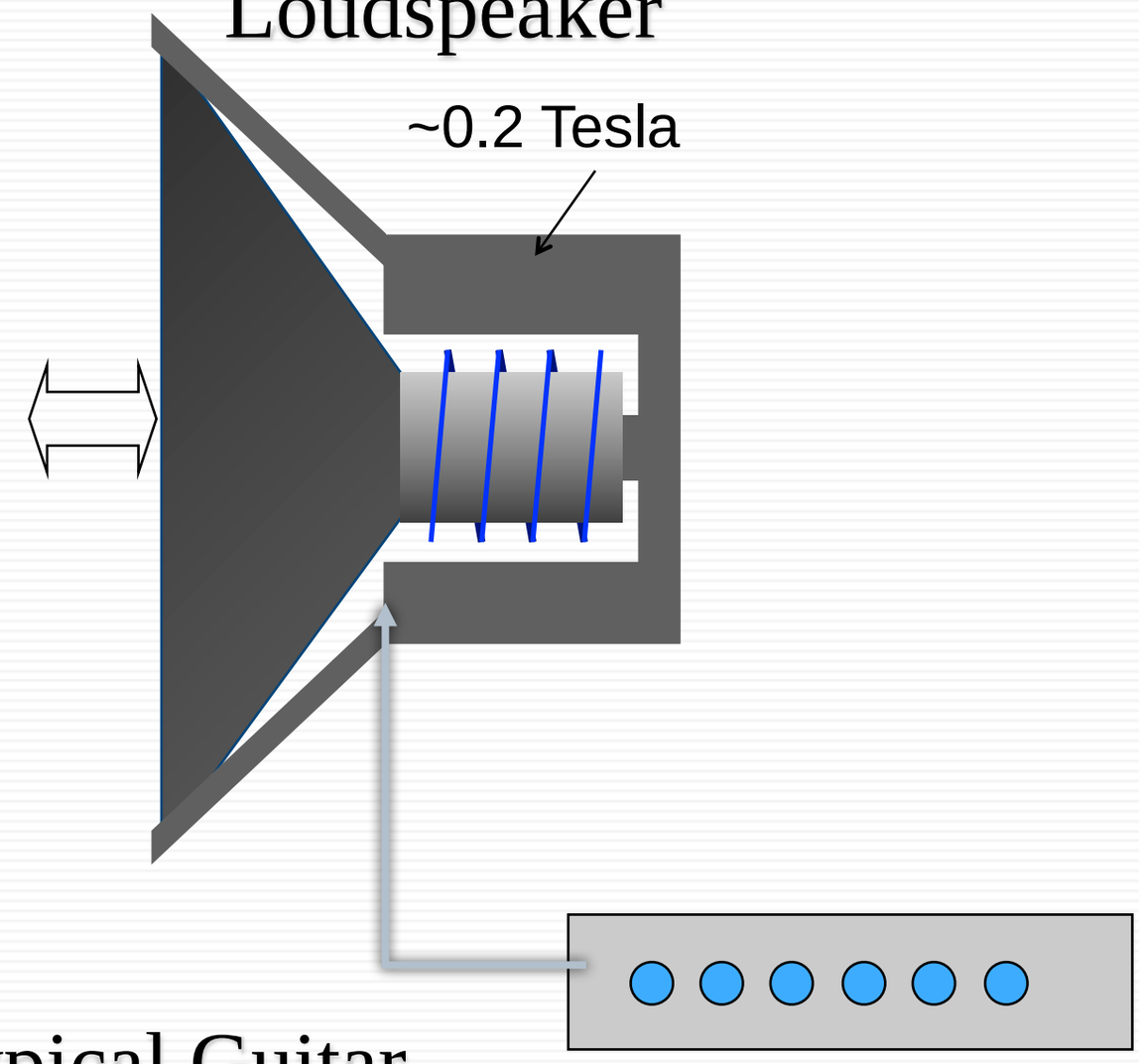


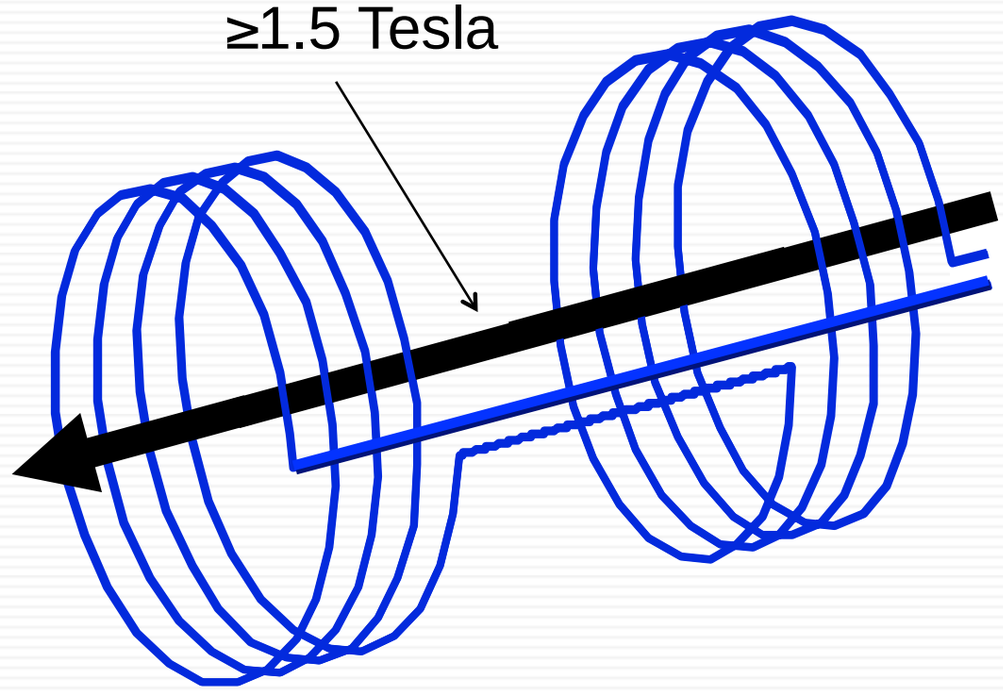
Typical Guitar Amplifier: **100** Watts

Why is MRI So Noisy?

Loudspeaker



MR Field Gradient



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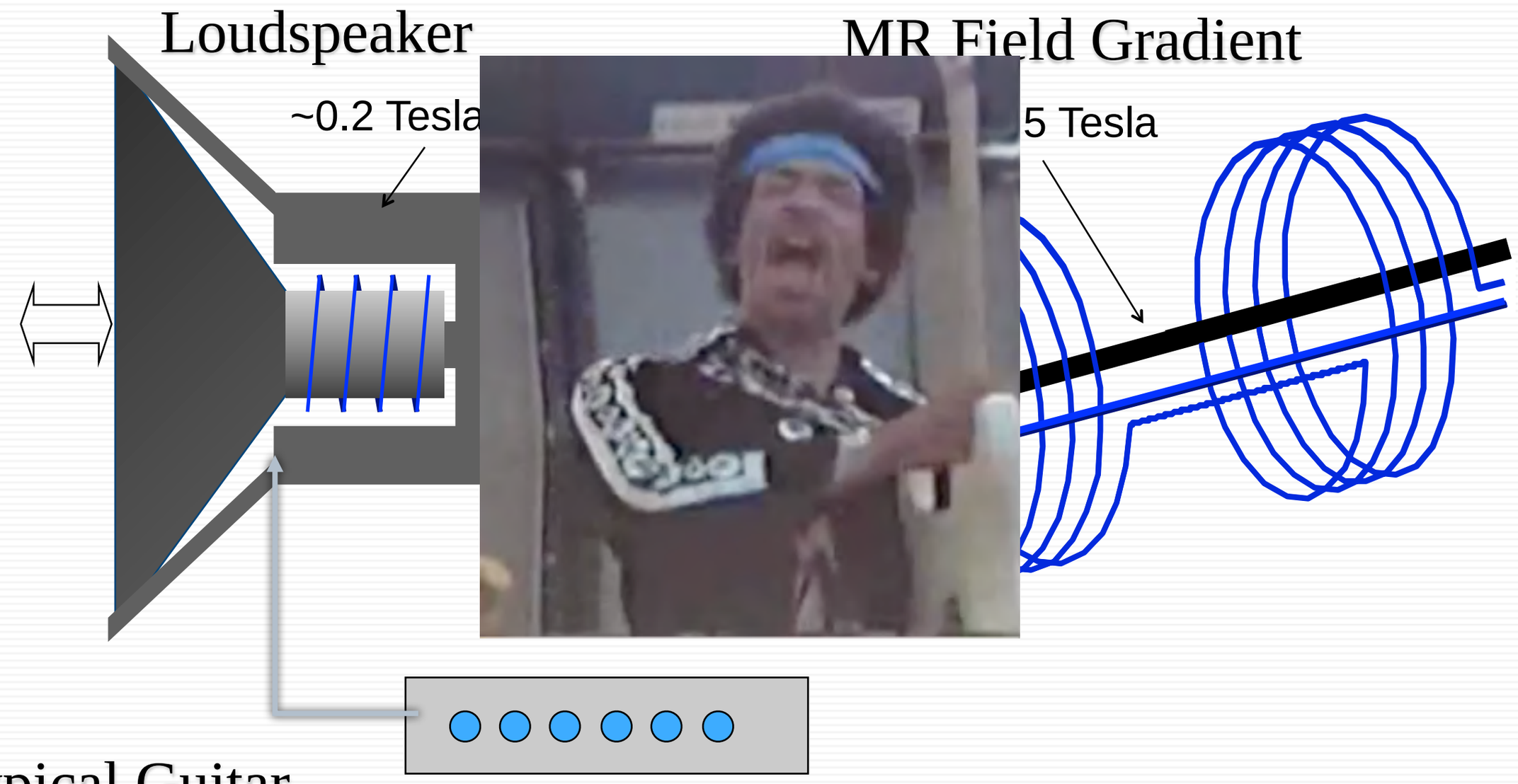

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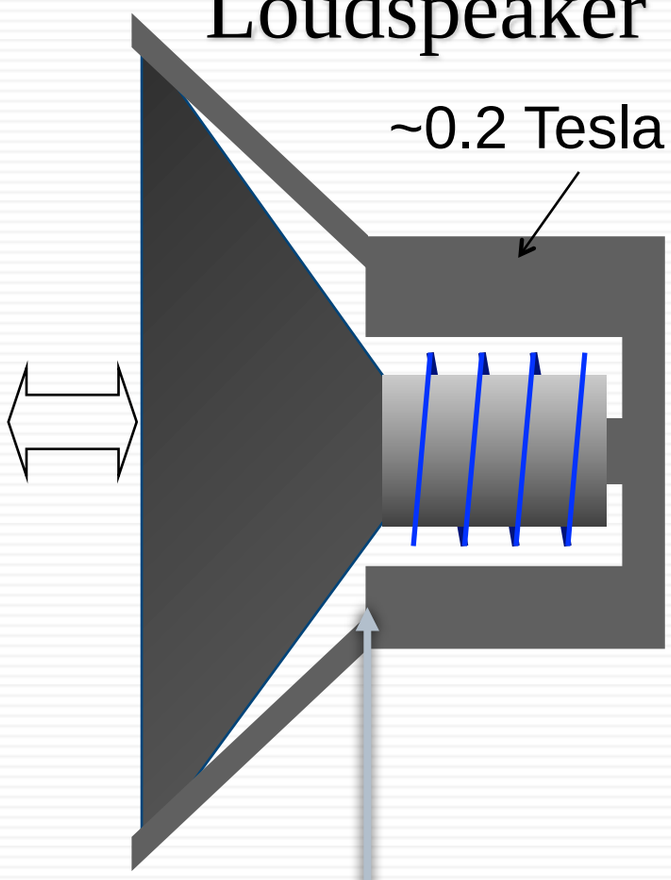
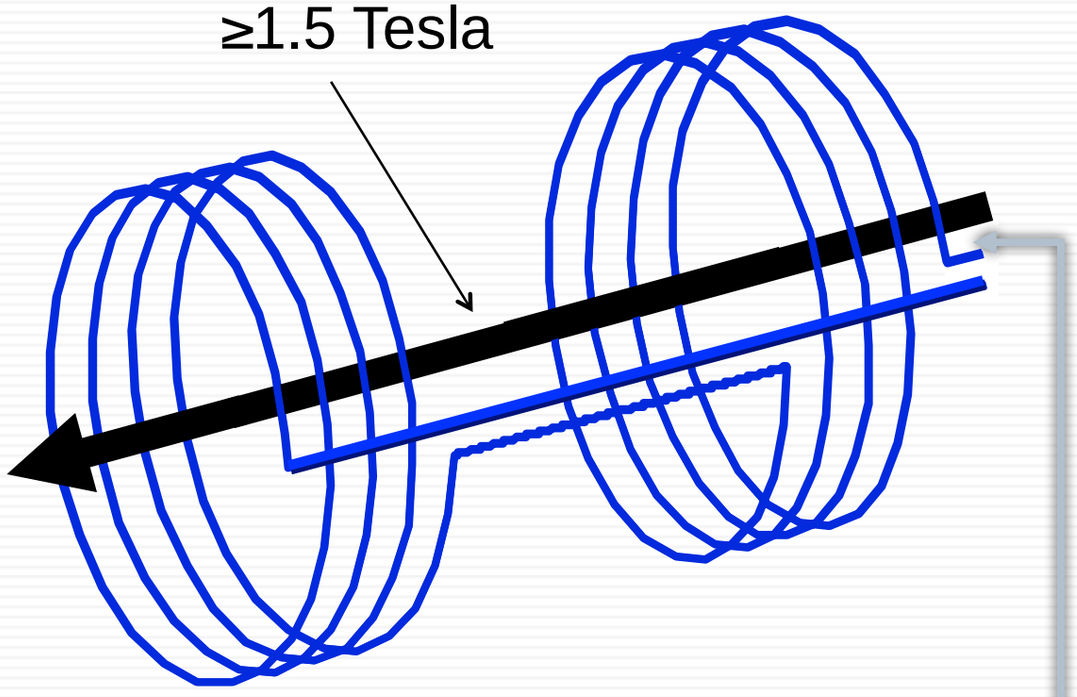
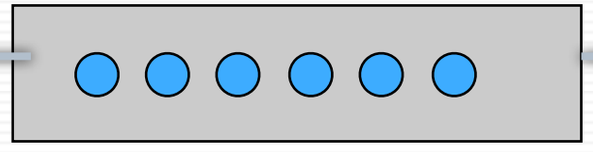


MR Field Gradient





Typical Guitar Amplifier: **100** Watts

Ultra-fast (echo-planar) gradient Amplifier: **865,000** Watts

The Plan

- The Magnetic Resonance Phenomenon & Contrast (30)
- Spatial Encoding (26)
- The "Pulse Sequence" Rules Everything (3)

Seventh Inning Stretch

- Fast Imaging (14)
- Functional MRI (18)
- Diffusion and Summary (9)
- Image Quality and Artifacts (48)

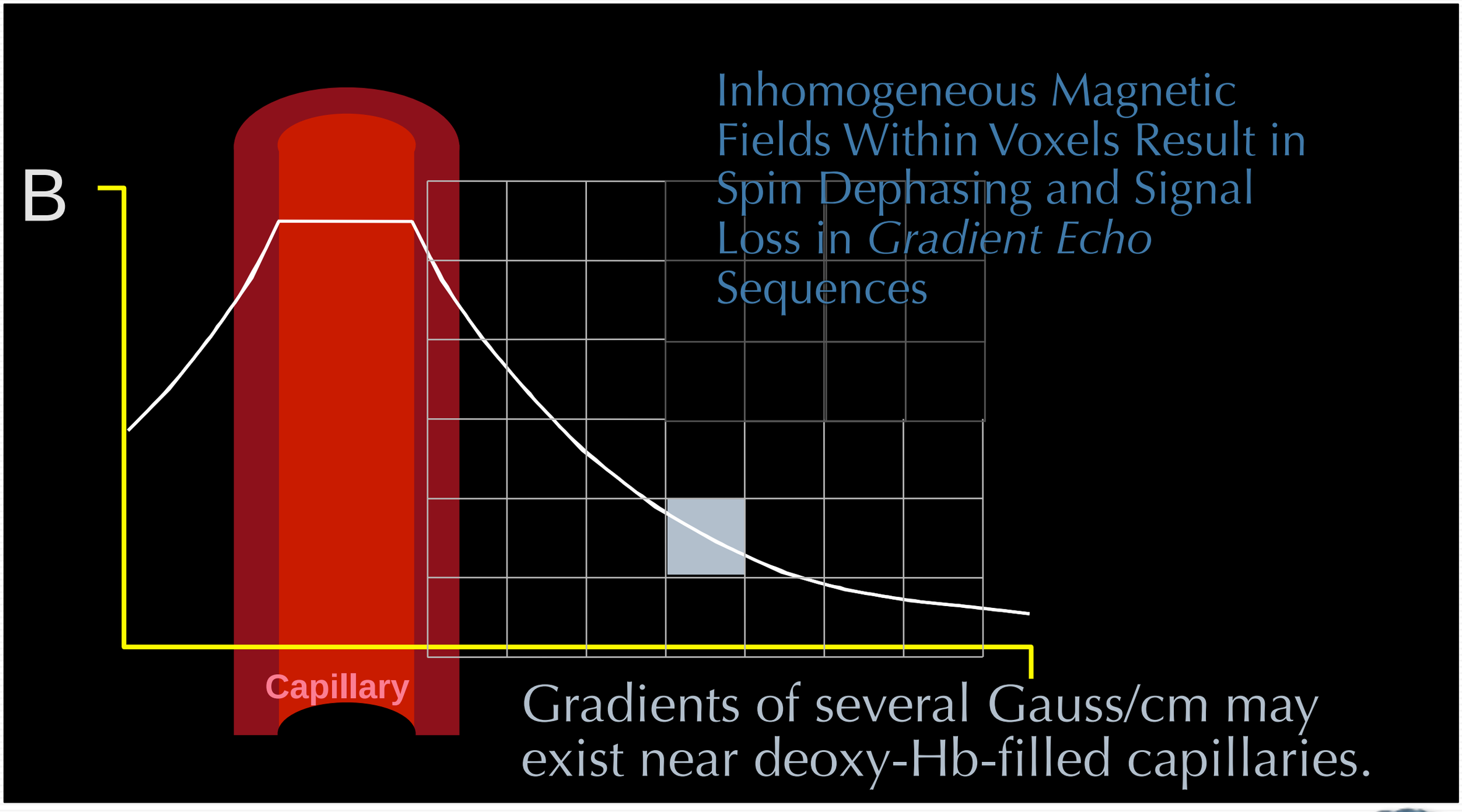
Brain "Activation" Leads to:

CBF	Increased	+ΔR1
CBV	Increased	+ΔR2 (C+)
O_2 Utilization	Increased slightly?	
Venous $[O_2]$	Increased	-ΔR2* ← "BOLD"
Glucose Utilization	Increased	? Lactate

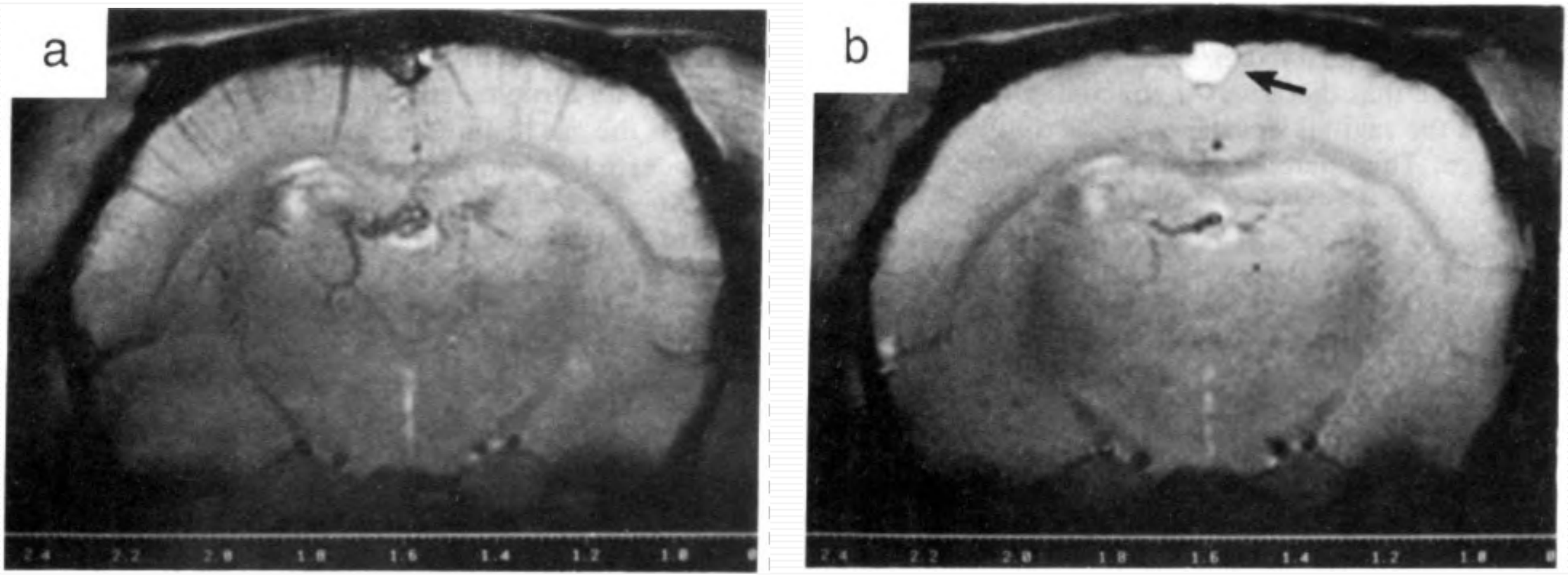
$R1 = 1/T1$

$R2 = 1/T2$

Signal Losses from Spin Dephasing



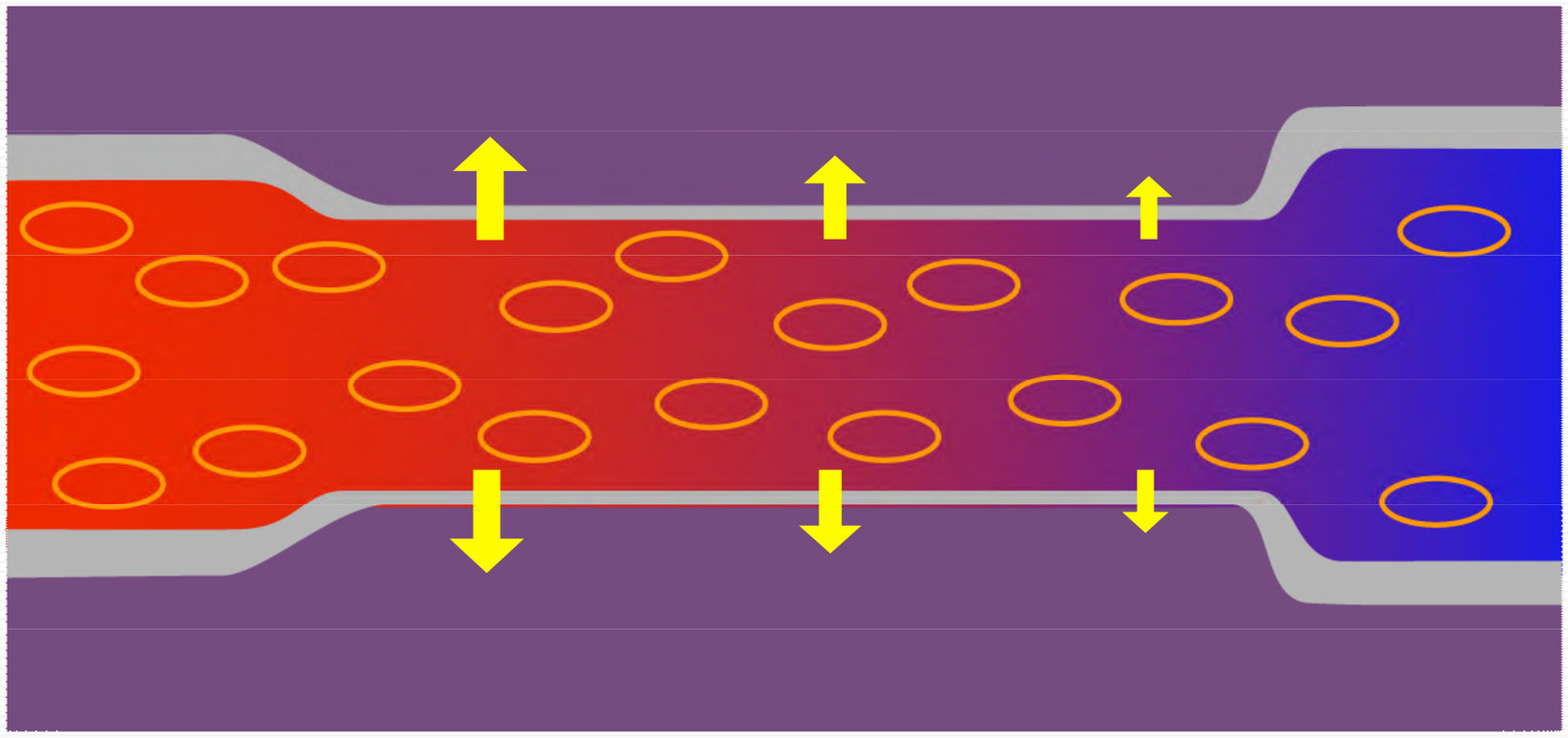
BOLD

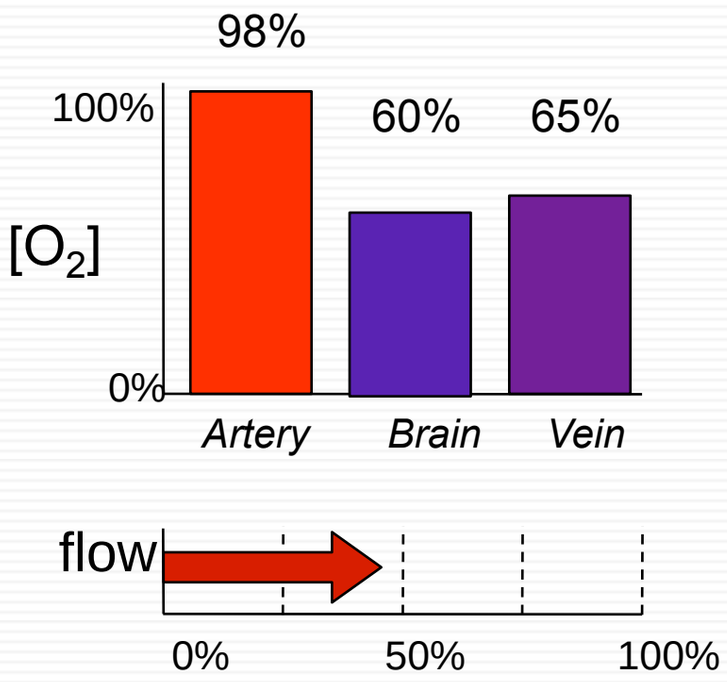


Effect of blood CO_2 level on BOLD contrast.
(*a*) Coronal slice brain image showing BOLD contrast from a rat anesthetized with urethane. The gas inspired was 100% O_2.
(*b*) The same brain but with 90% O_2/10%CO_2 as the gas inspired. BOLD contrast is greatly reduced.

S Ogawa, *et al.*,
PNAS, **87**(24):9868,1990

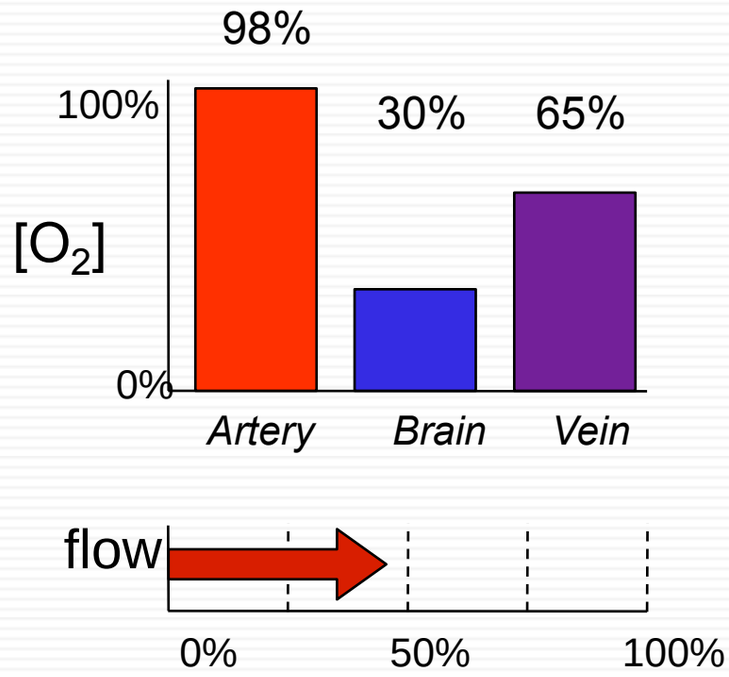
Why Does Venous O_2 Increase? (1)





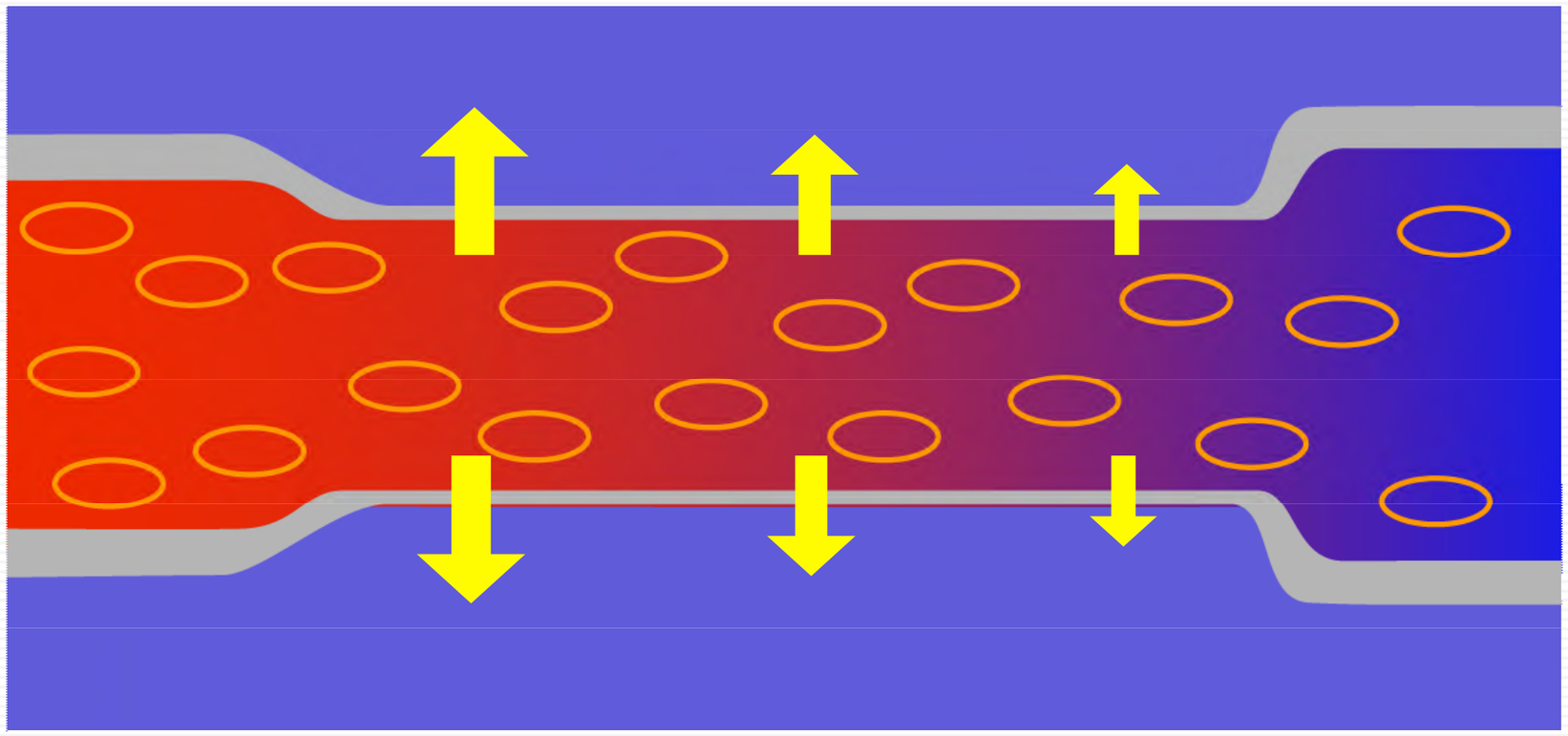
Under normal conditions oxygen diffuses down its concentration gradient from the capillary to the brain parenchyma

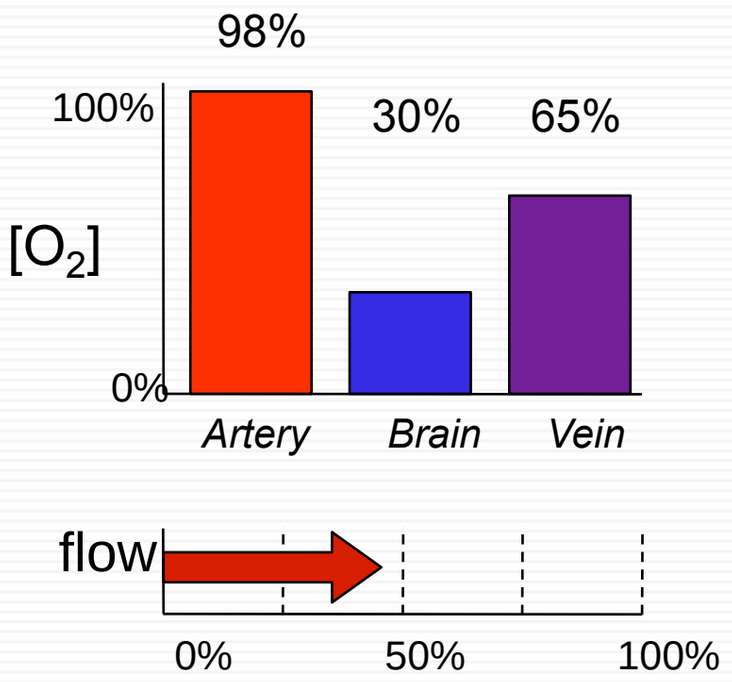
Why Does Venous O_2 Increase? (2)



As the brain becomes more active, the oxygen consumption increases, increasing the transluminal oxygen gradient.

Why Does Venous O_2 Increase? (2)





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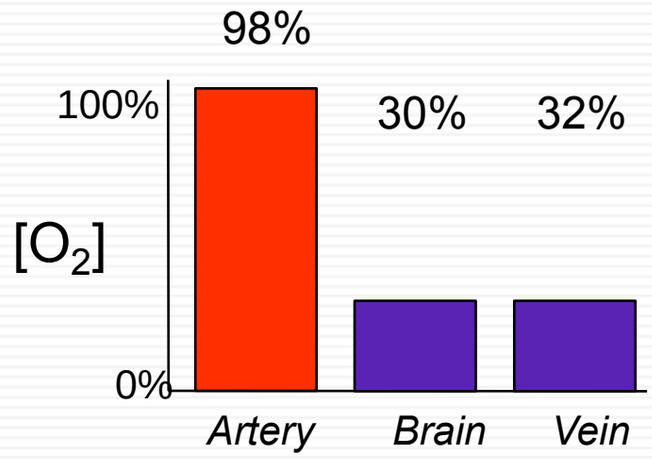
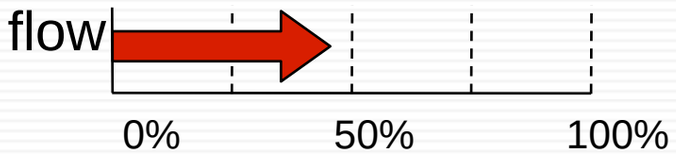
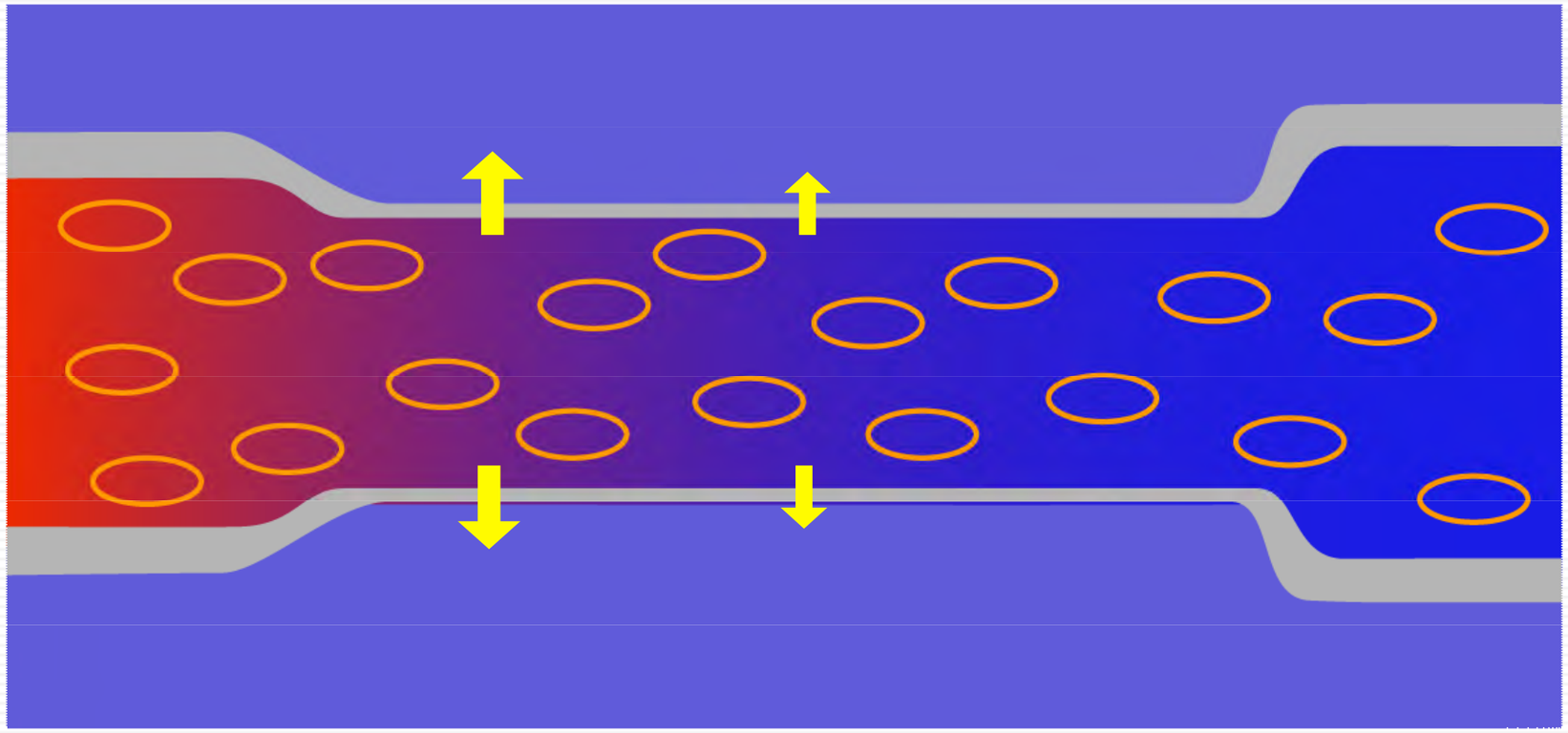
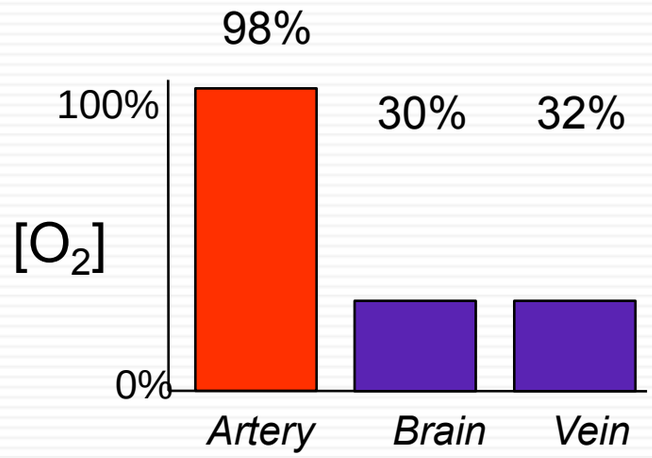


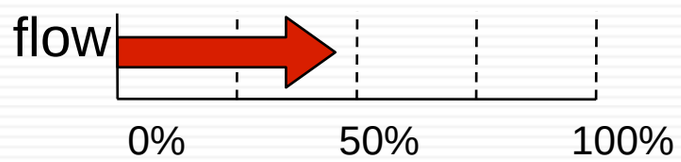


As oxygen flows across the capillary lumen it is depleted in the capillary and no further oxygen can be delivered

Why Does Venous O_2 Increase? (3)







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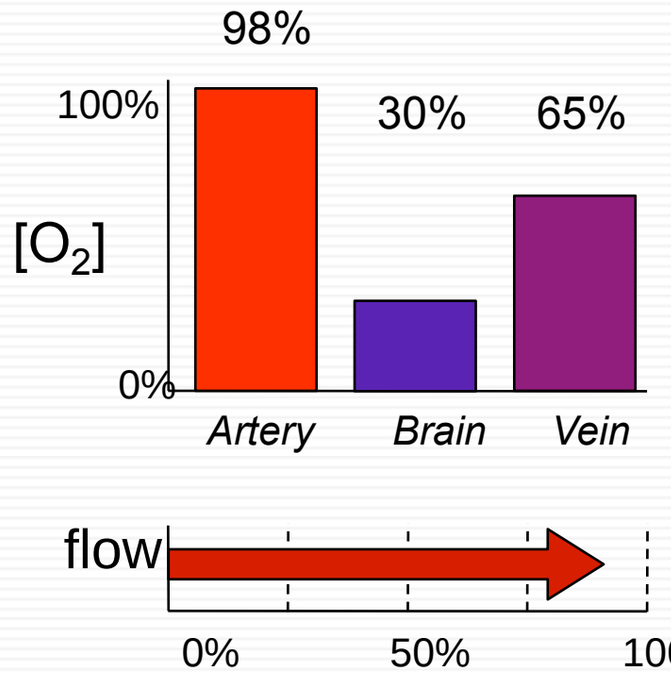
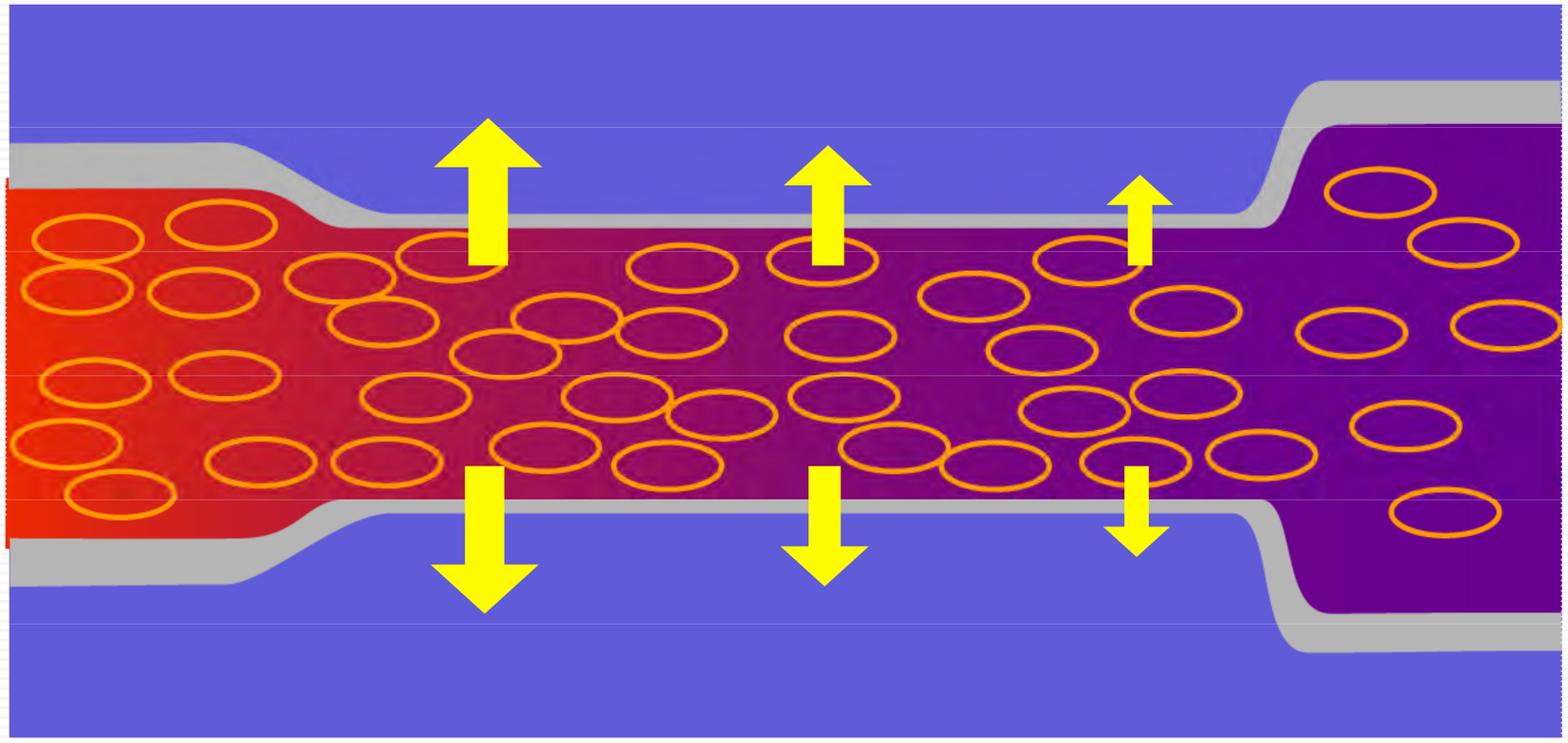
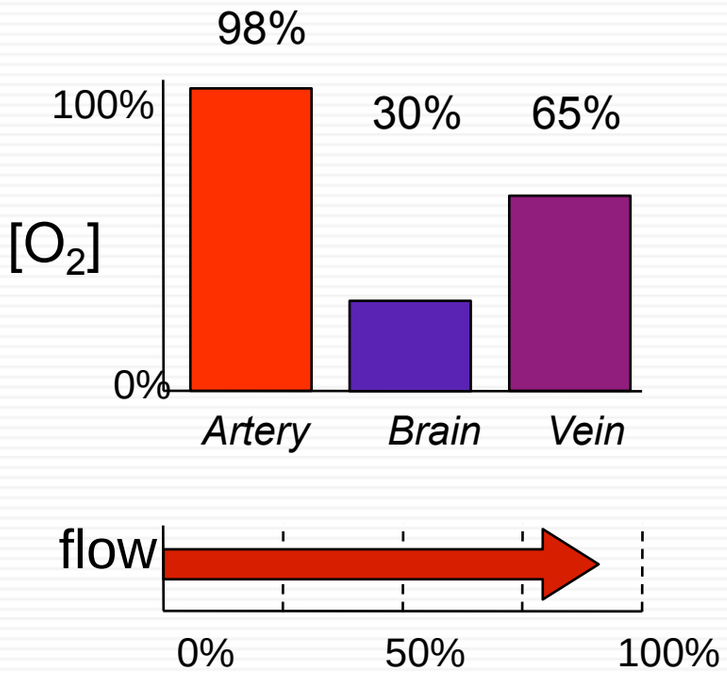
Why Does Venous O_2 Increase? (4)



The vascular system responds by increasing blood flow so that more oxygenated blood is available throughout the capillary

Why Does Venous O_2 Increase? (4)





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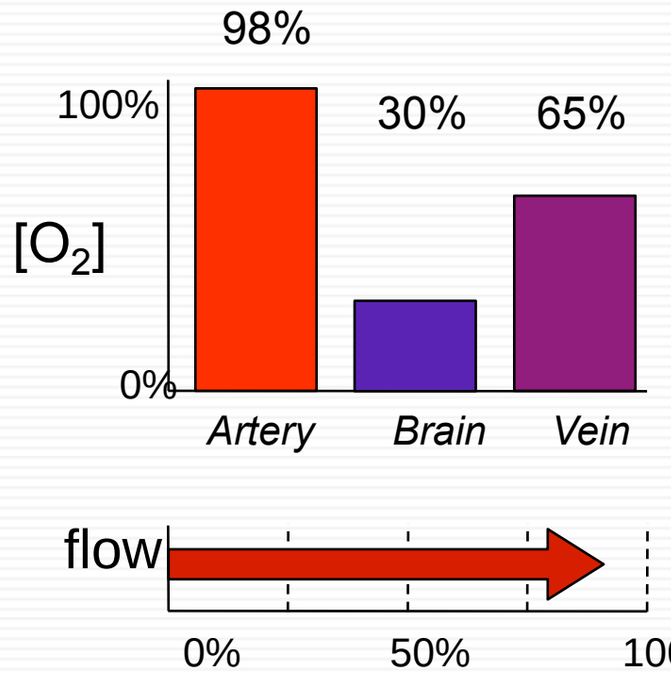
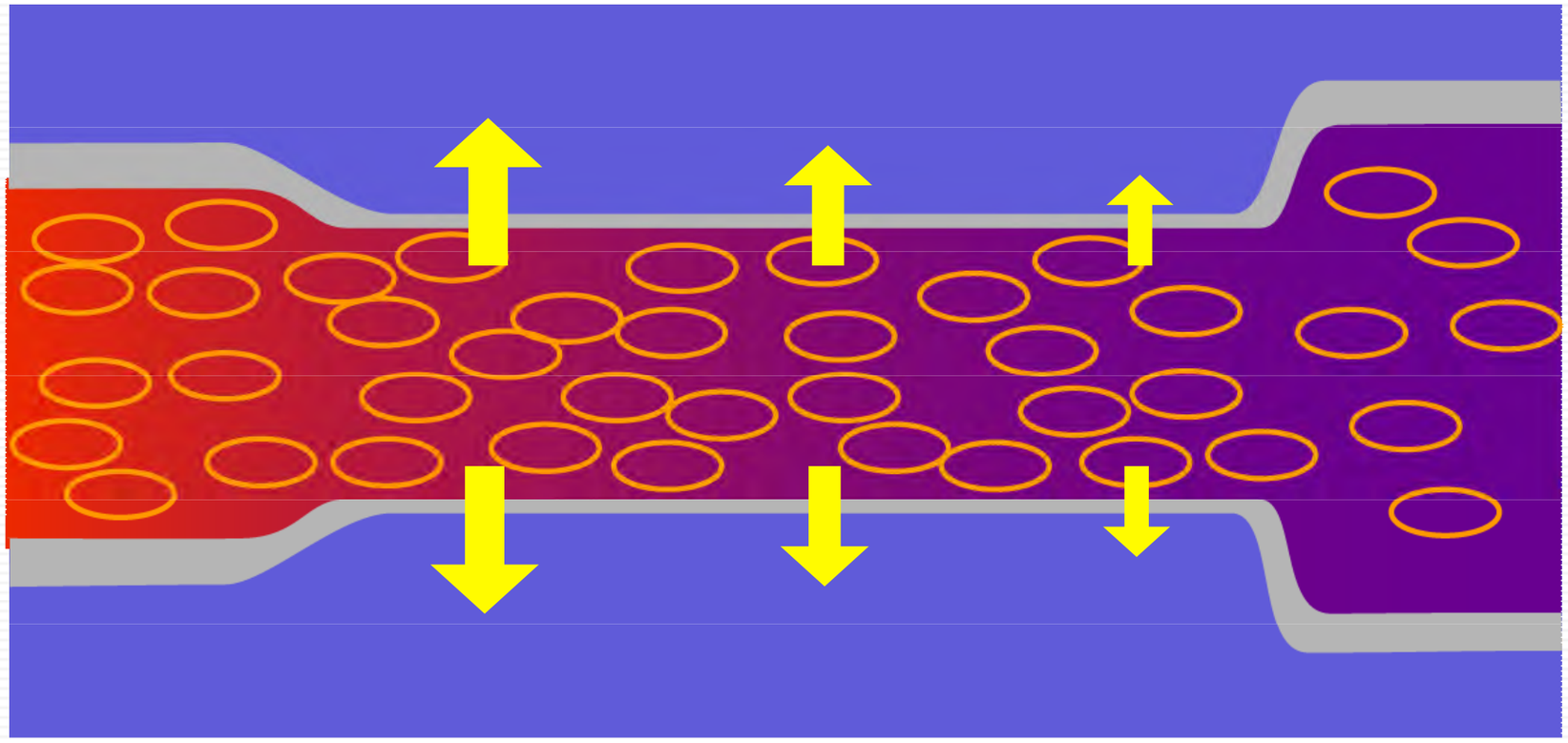
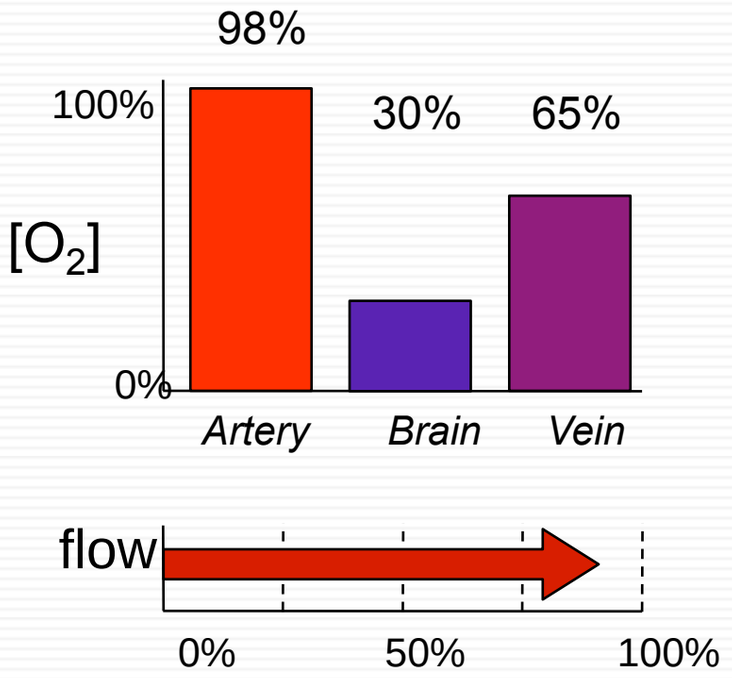


Because the blood flow is increased more oxygenated blood passes into the venous end of the capillary

Why Does Venous O_2 Increase? (5)





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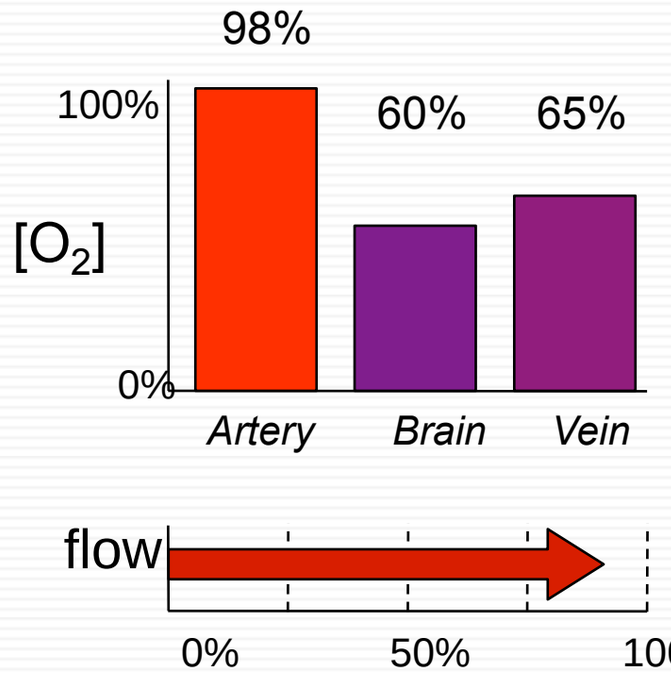

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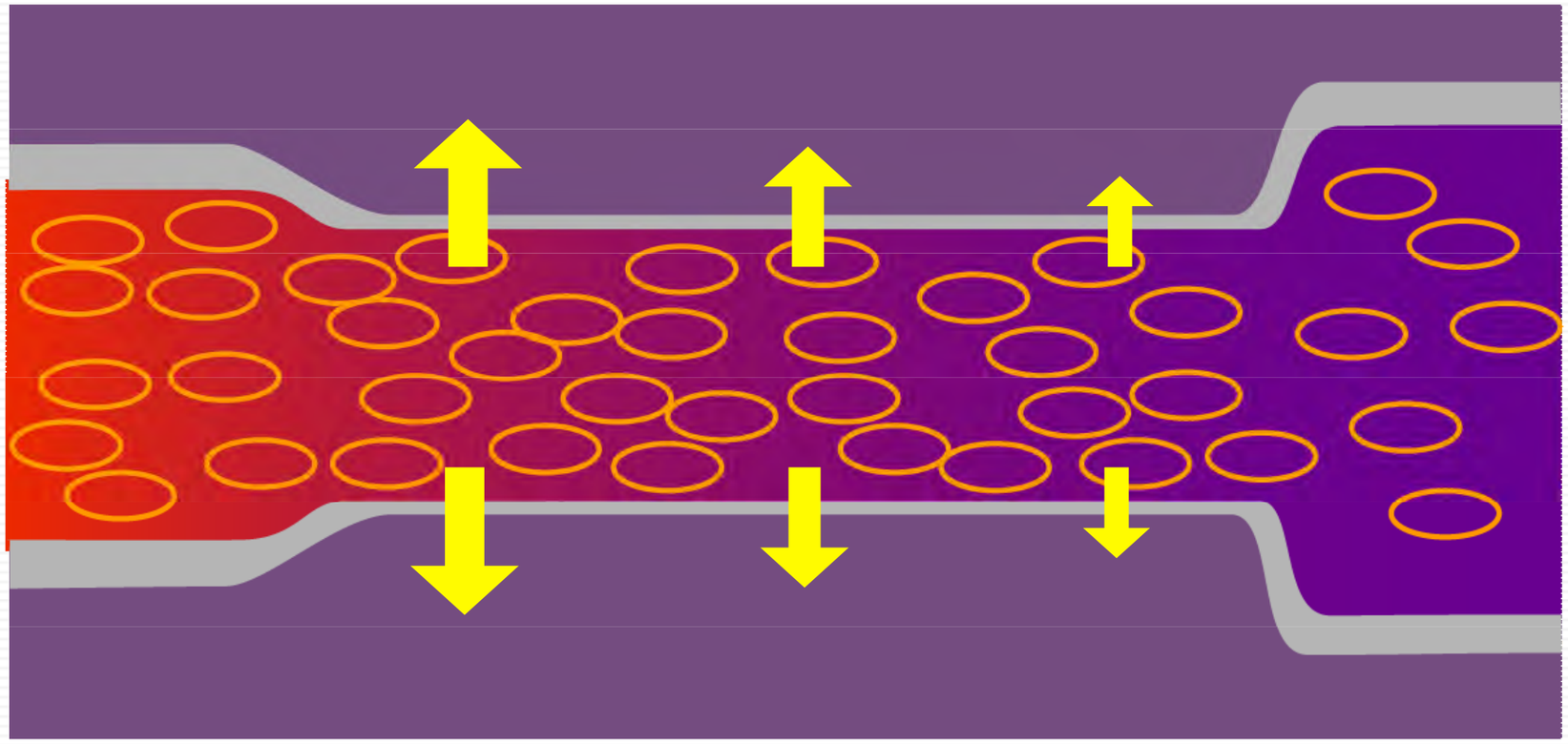

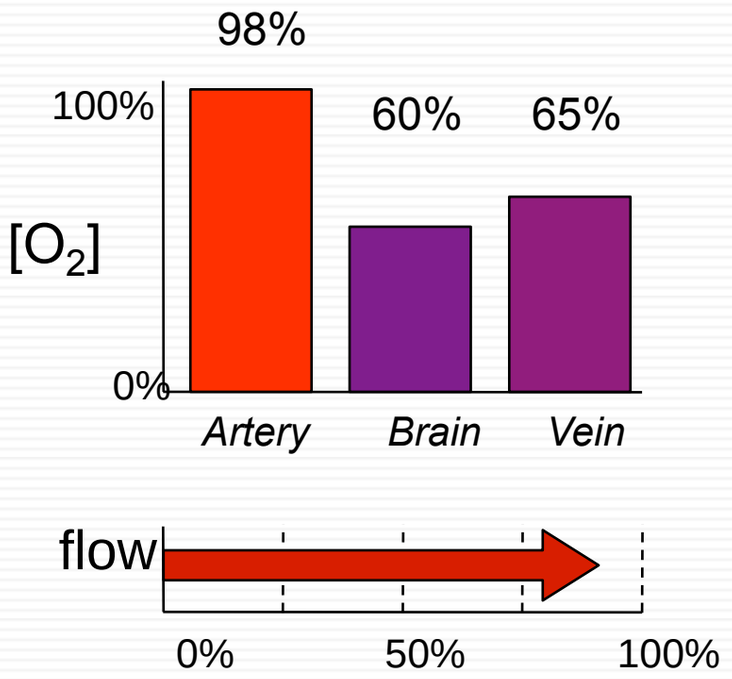

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BOLD Contrast & Field Strength

- BOLD Contrast arises from susceptibiity differences
- The absolute field distortion (from BOLD) is proportional to the magnetic field strength
- The absolute change in MRI signal is proportional to *both* the field distortion and the signal strength.

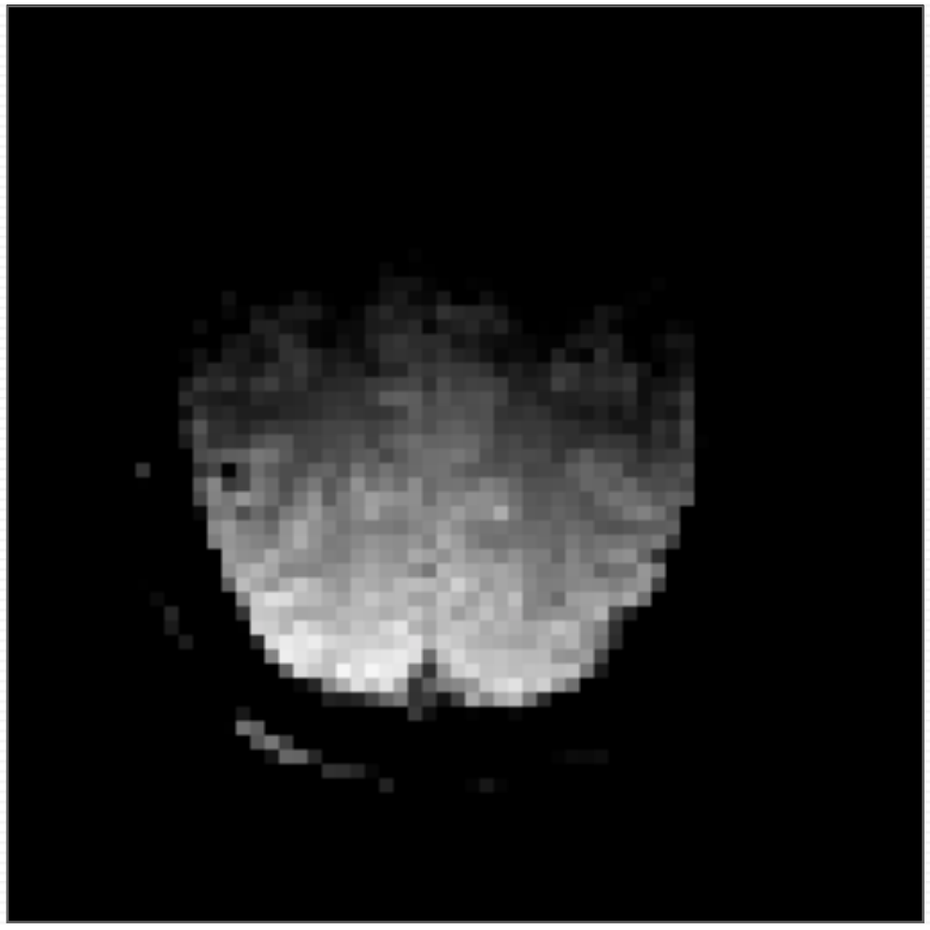
$$\text{BOLD } \textit{should} \text{ go as } k\mathrm{B}_0^2$$

fMRI

explores intensity variations in MR signal

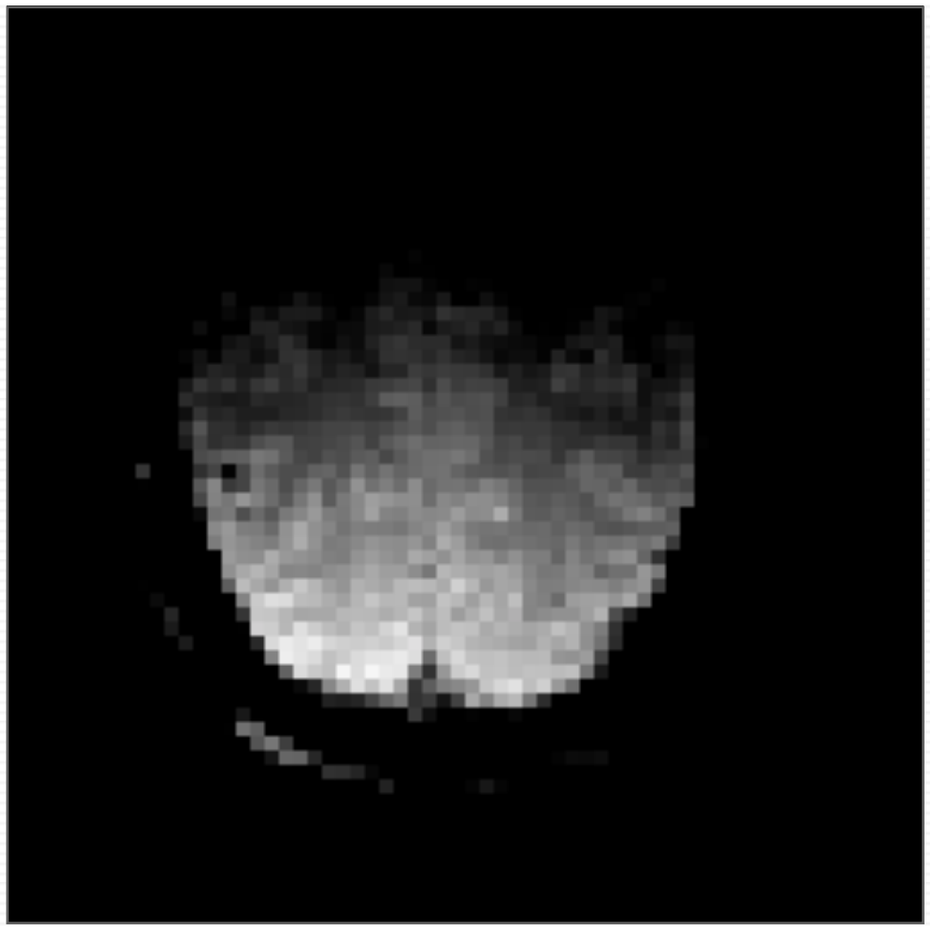
fMRI

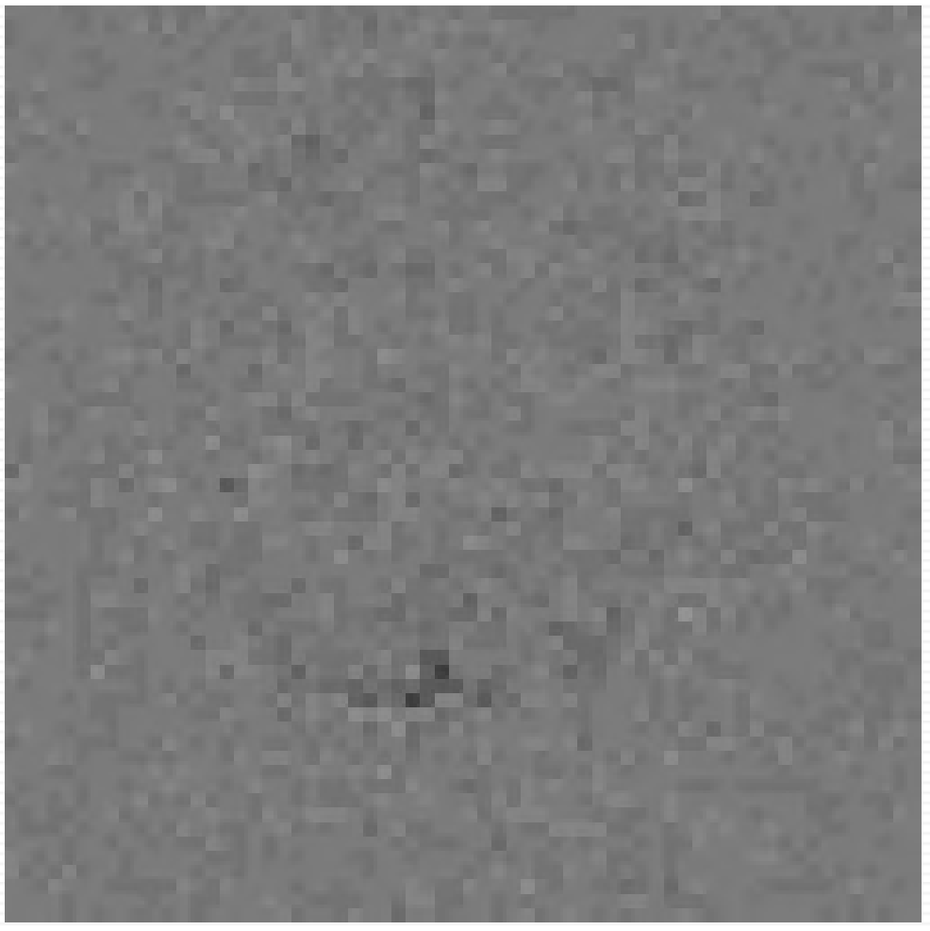
explores intensity variations in MR signal



fMRI

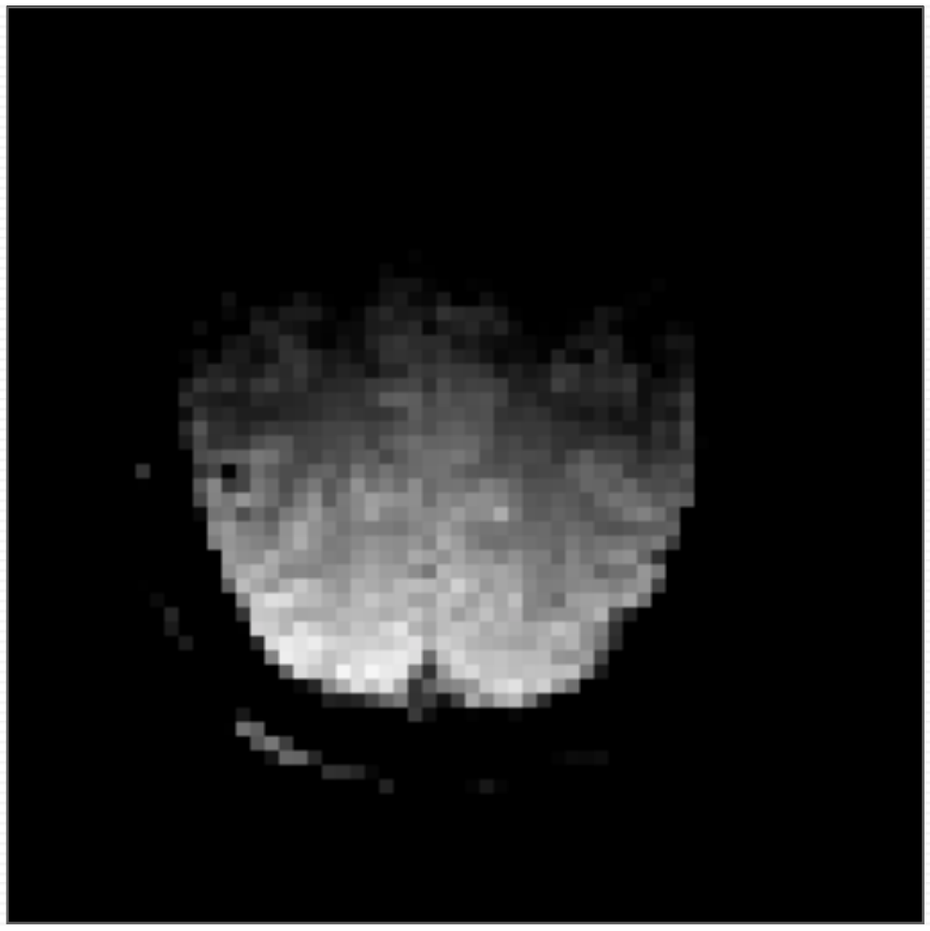
explores intensity variations in MR signal

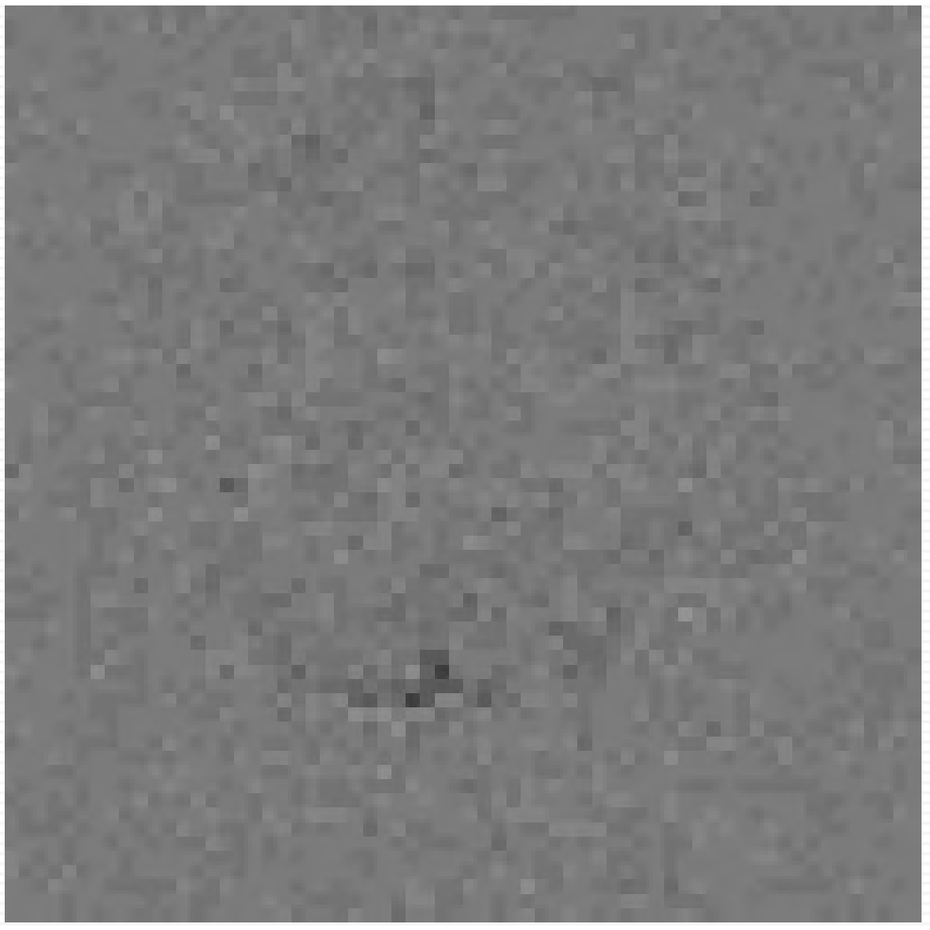




fMRI

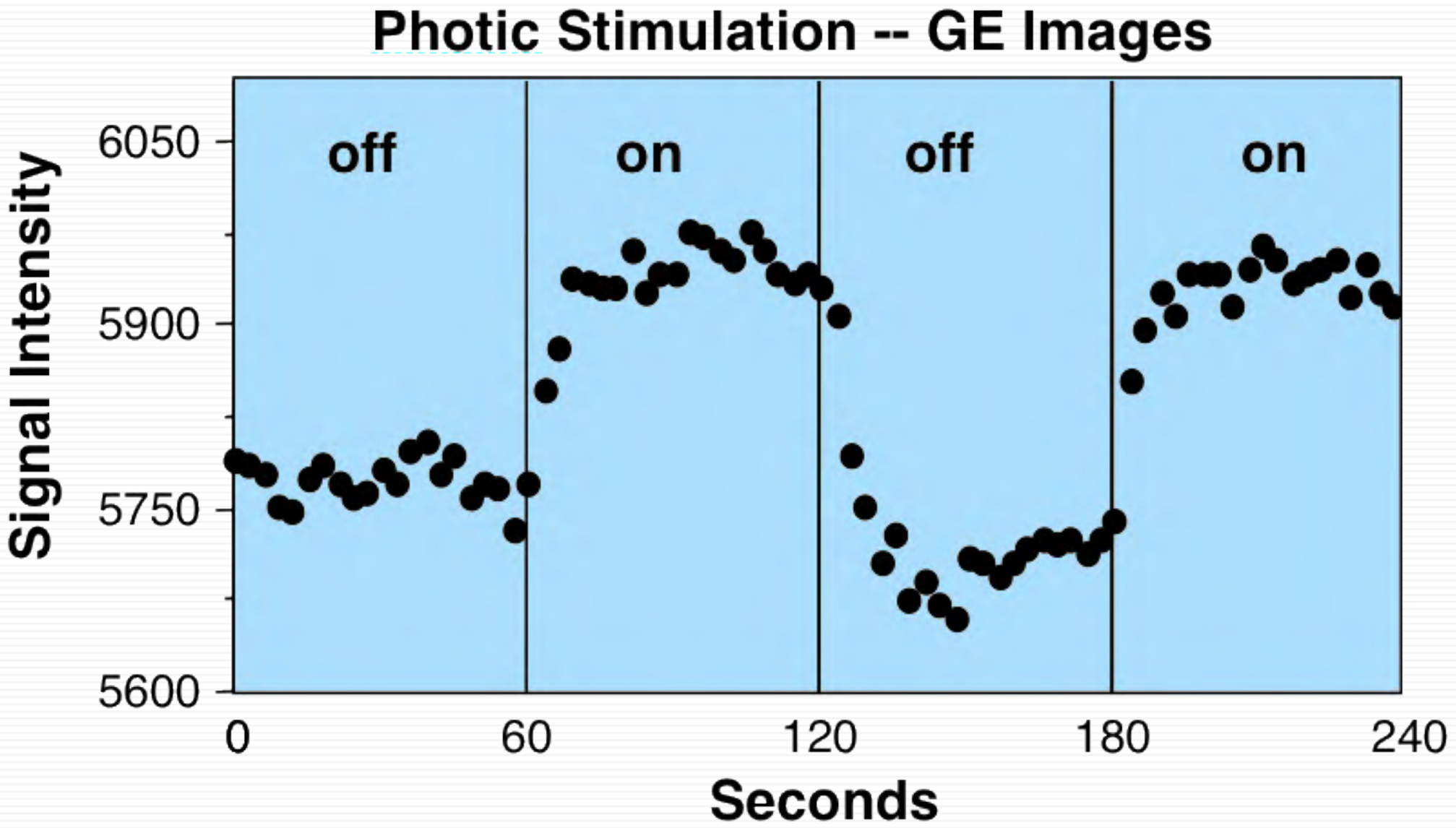
explores intensity variations in MR signal





intensity variations reflect venous [O2]

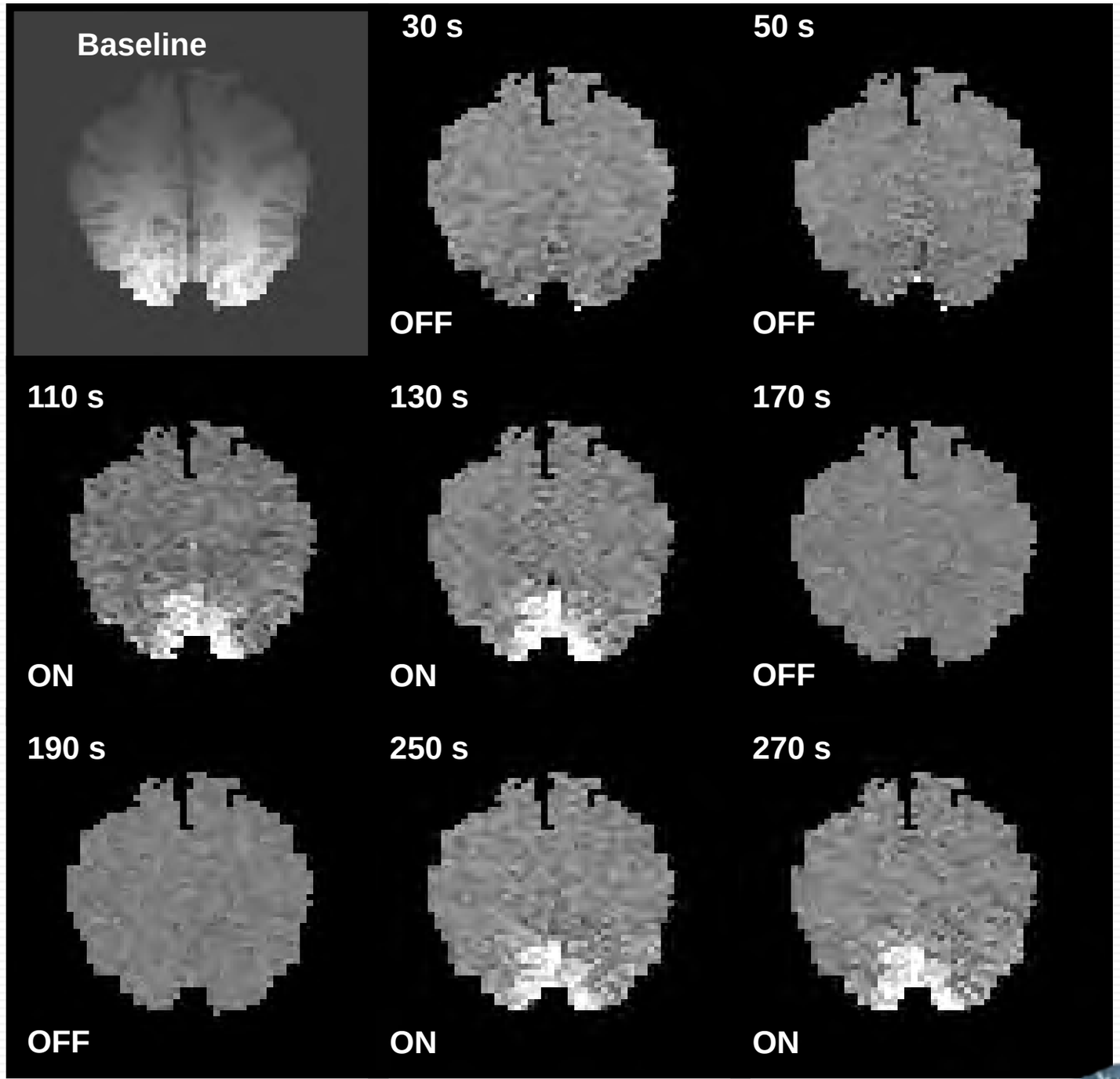
Gradient-Recalled Echo



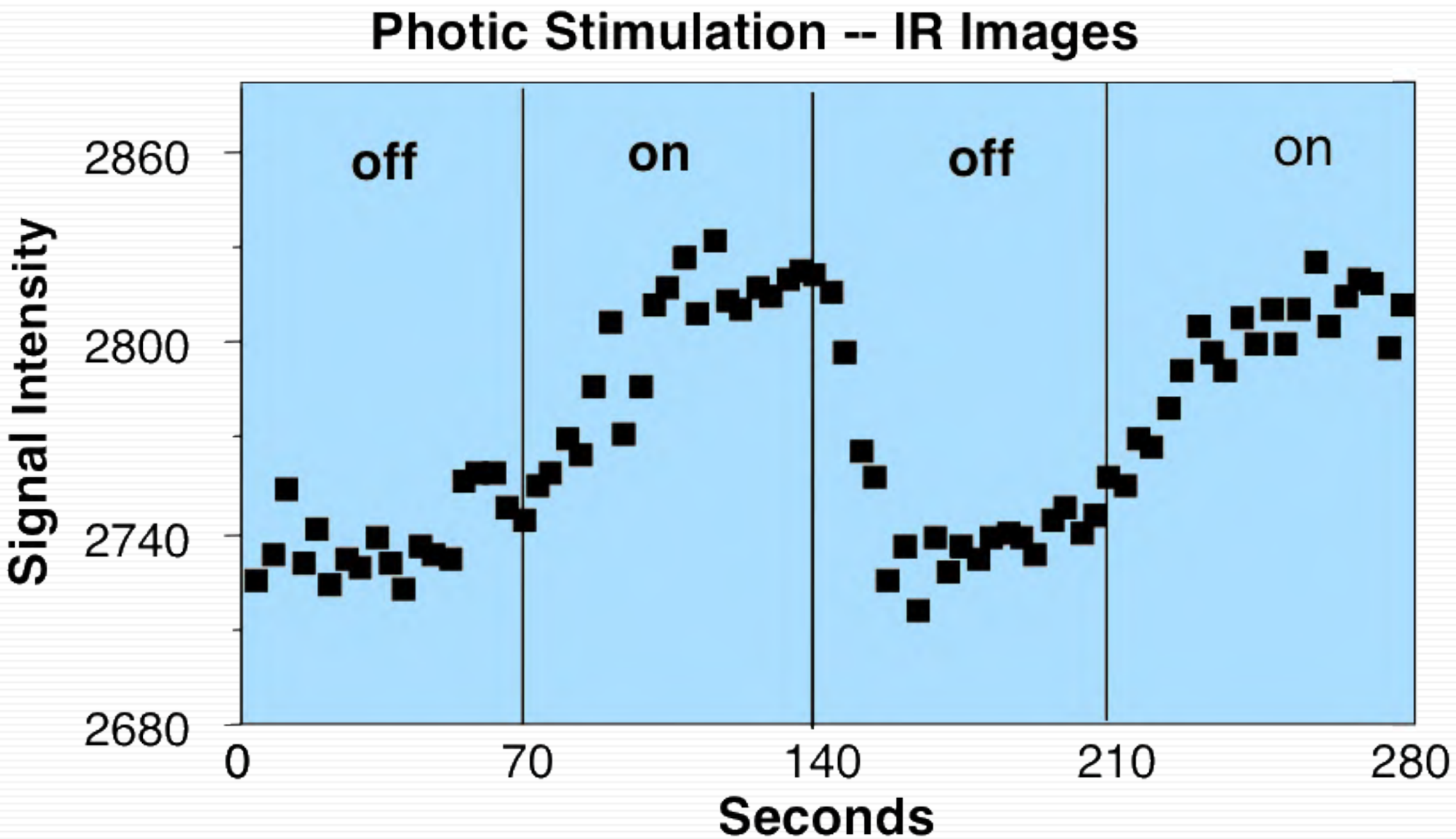
Ken Kwong

Ken Kwong

Inversion Recovery
TE=42 TR=3000
TI = 1100
Thickness=10

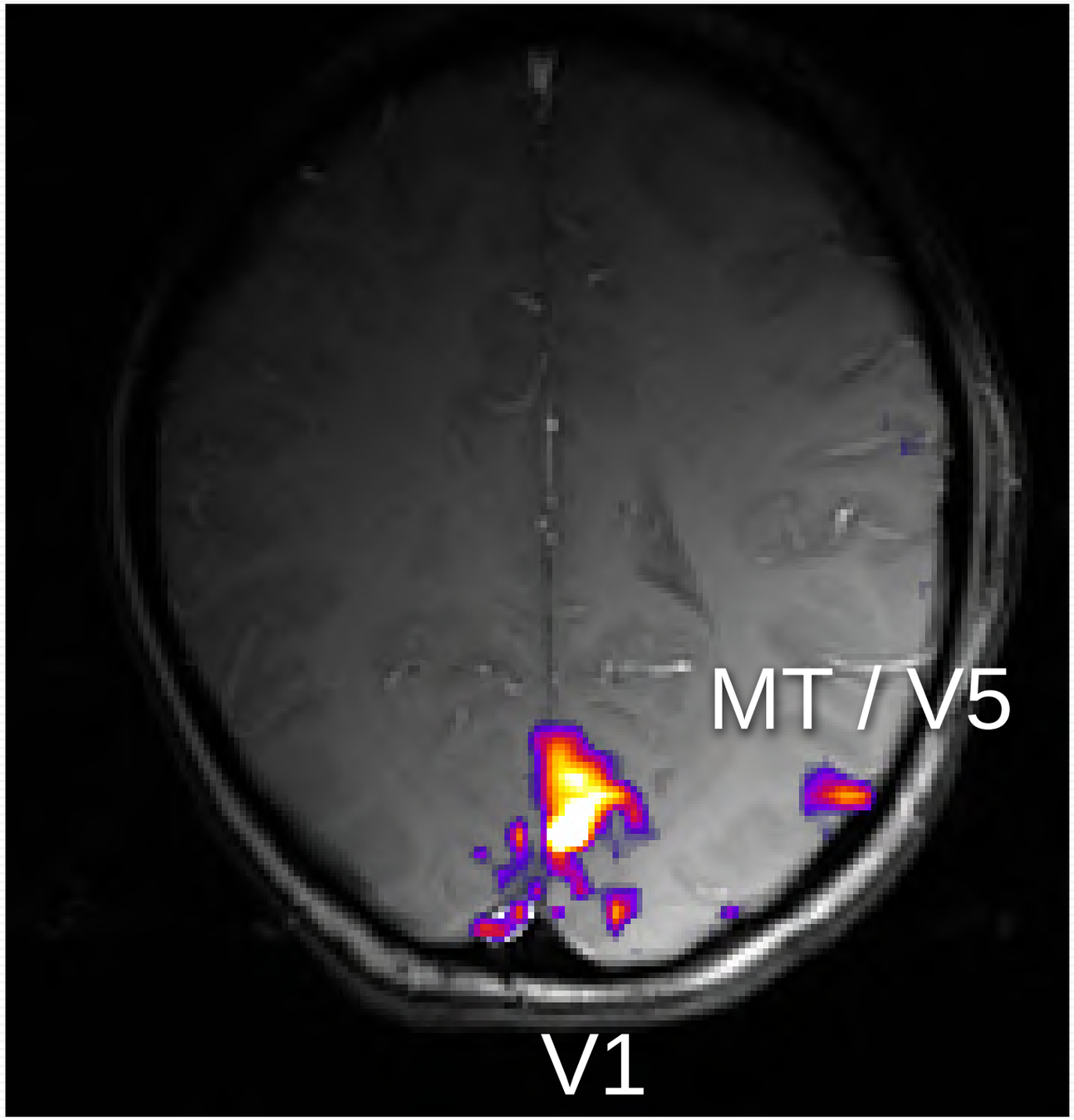


Inversion Recovery

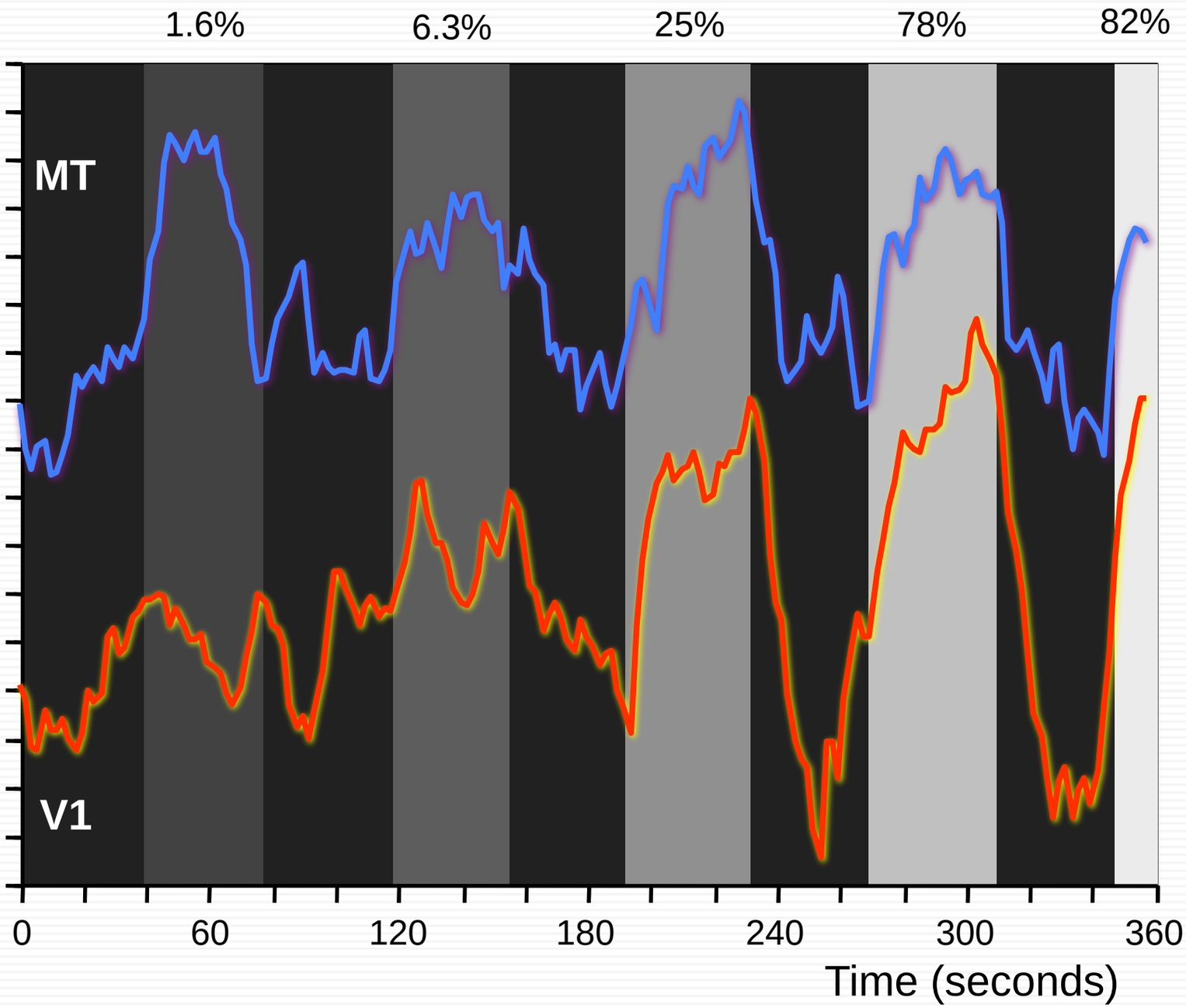


Ken Kwong

Activation with Moving Visual Stimuli

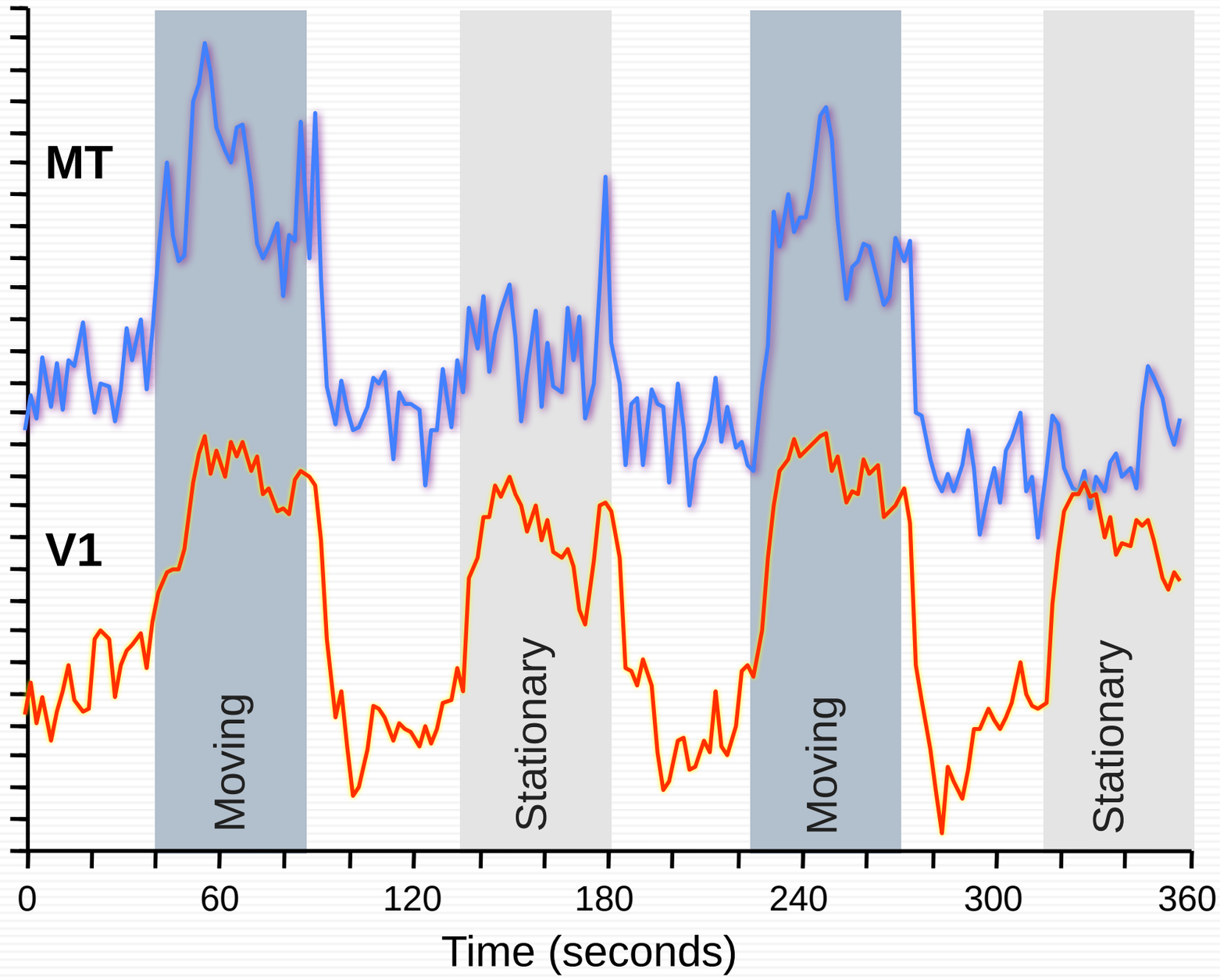


Contrast Response Test



From R. Tootell

Motion Sensitivity Test

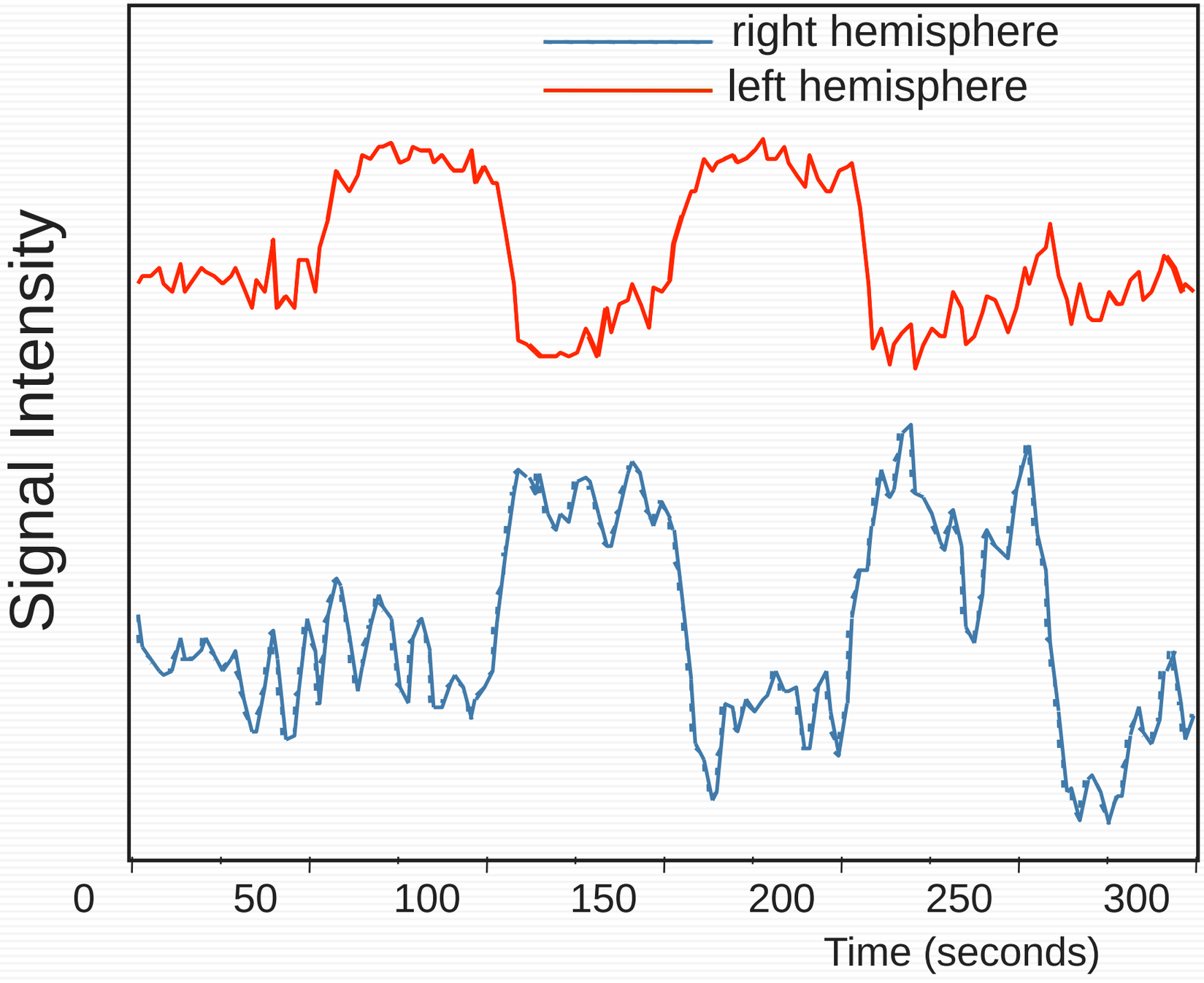


NeuroImaging Training Program

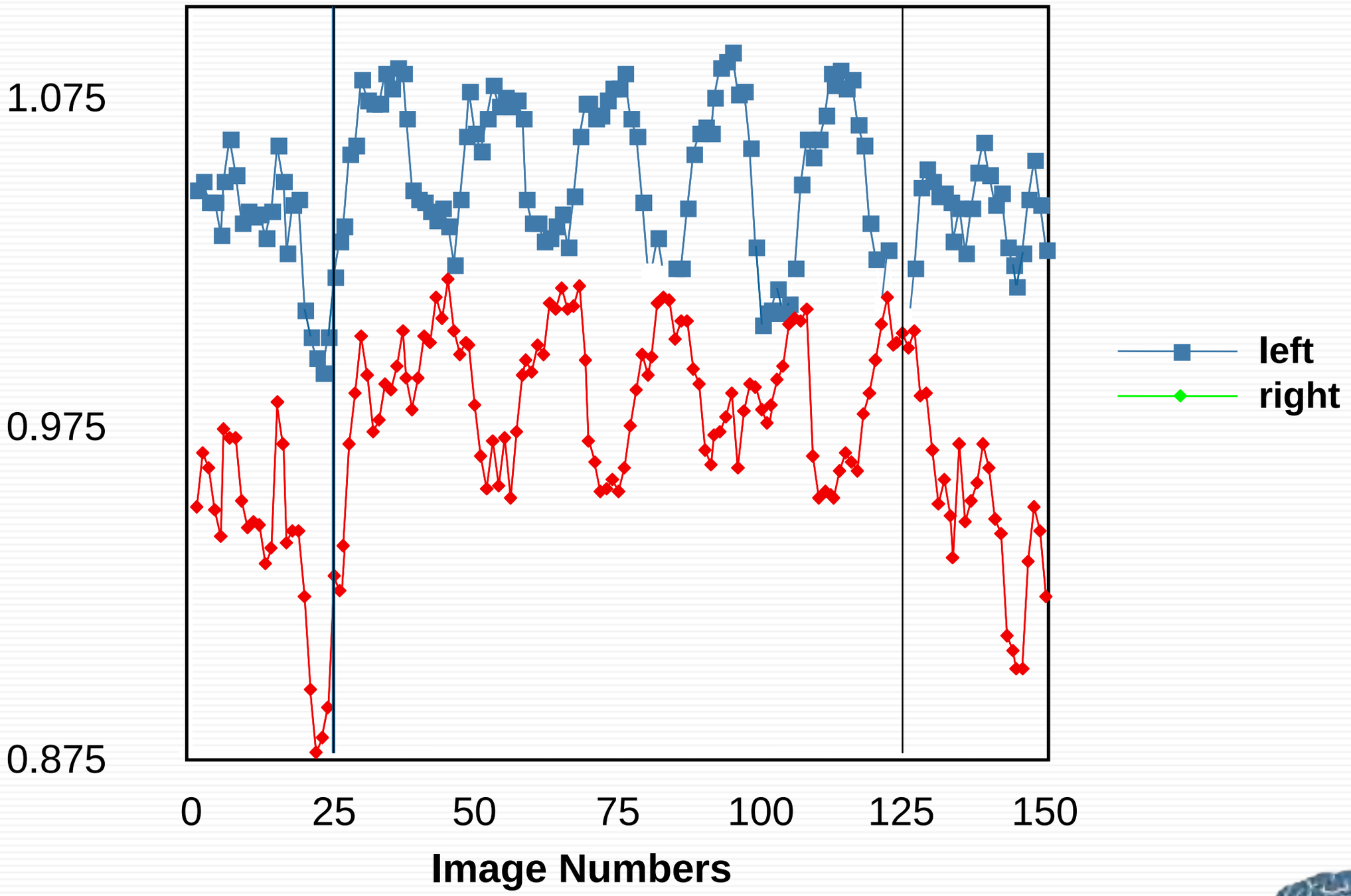
www.brainmapping.org

From R. Tootell Center for Cognitive Neuroscience

Hemifield Alternation



Hemifield Alternation 20 seconds



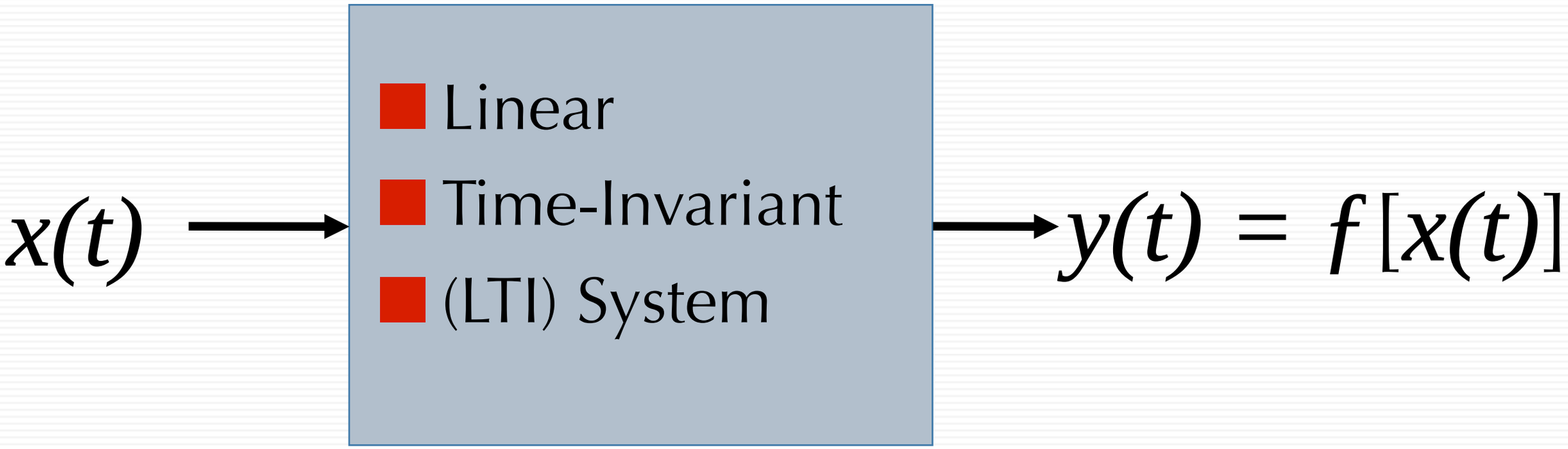
Neurovascular Coupling and *f*MRI latency

Pre-capillary Sphincters

Hydraulic Delay

Potassium Slow Potentials?
VIP release?
NO release?

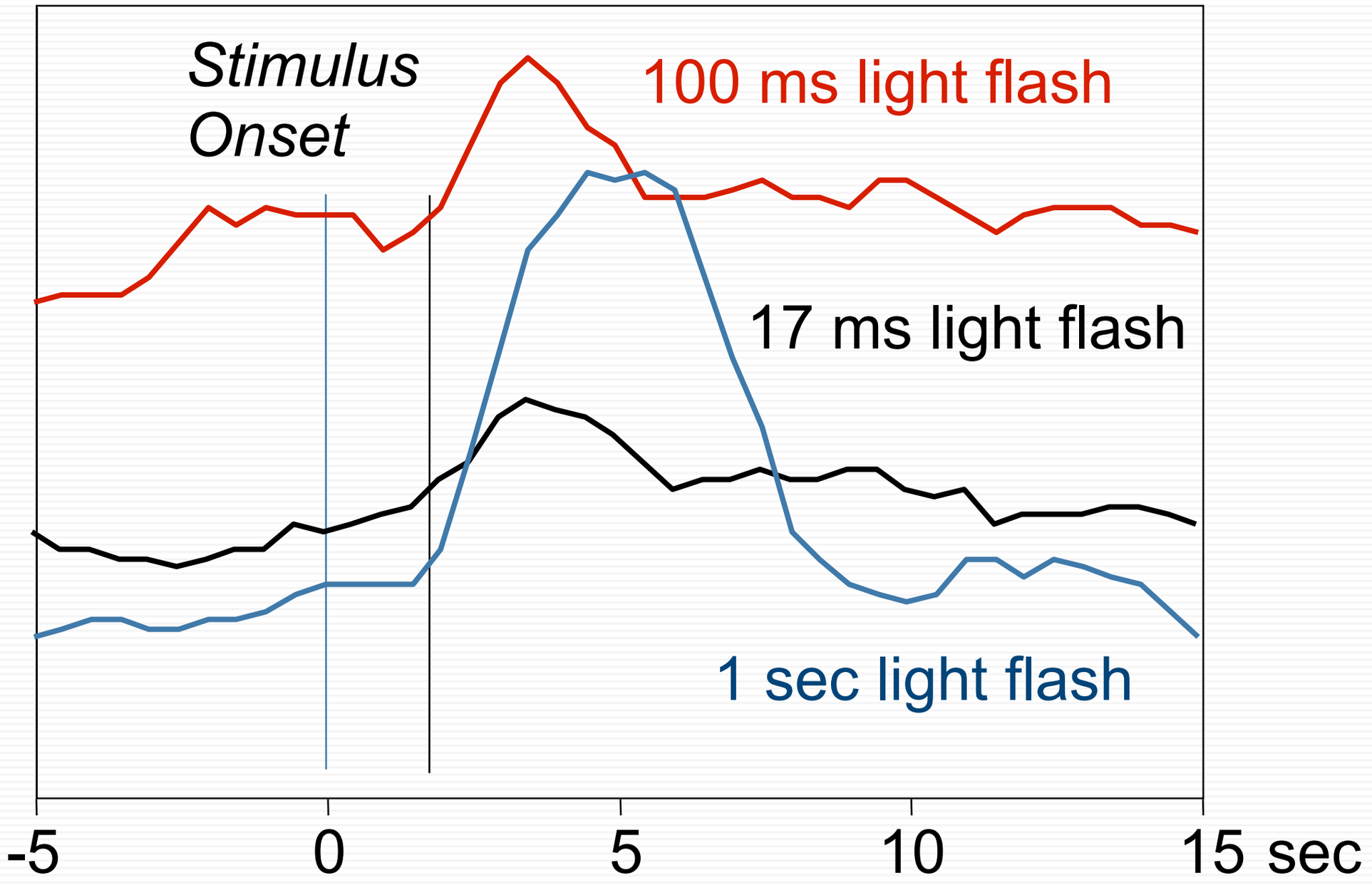
Linear Systems Approach



In an LTI system, given two inputs A & B:

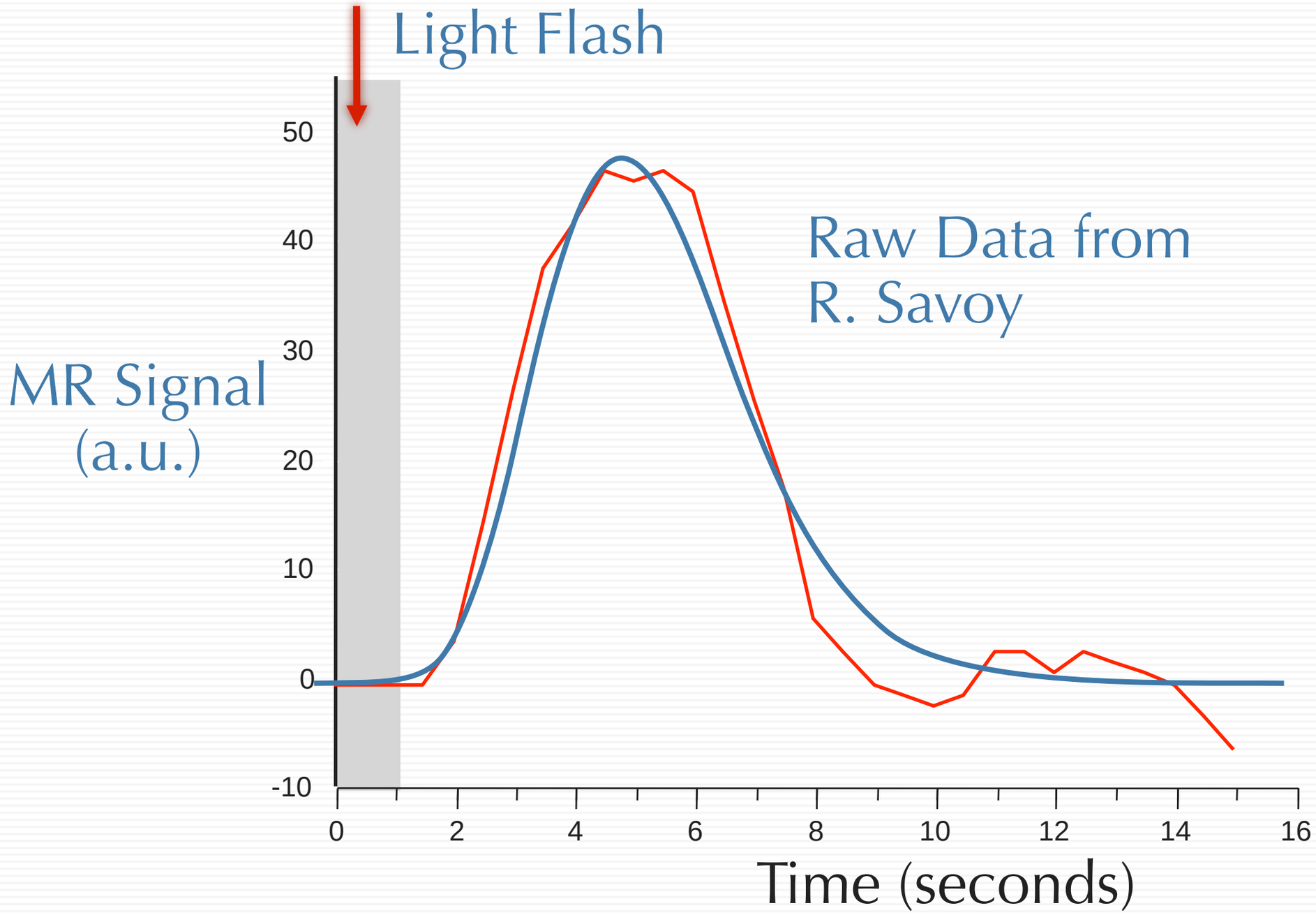
$$f(A + B) = f(A) + f(B)$$

Response Latency vs. Stimulus Duration
Average of 10 recordings

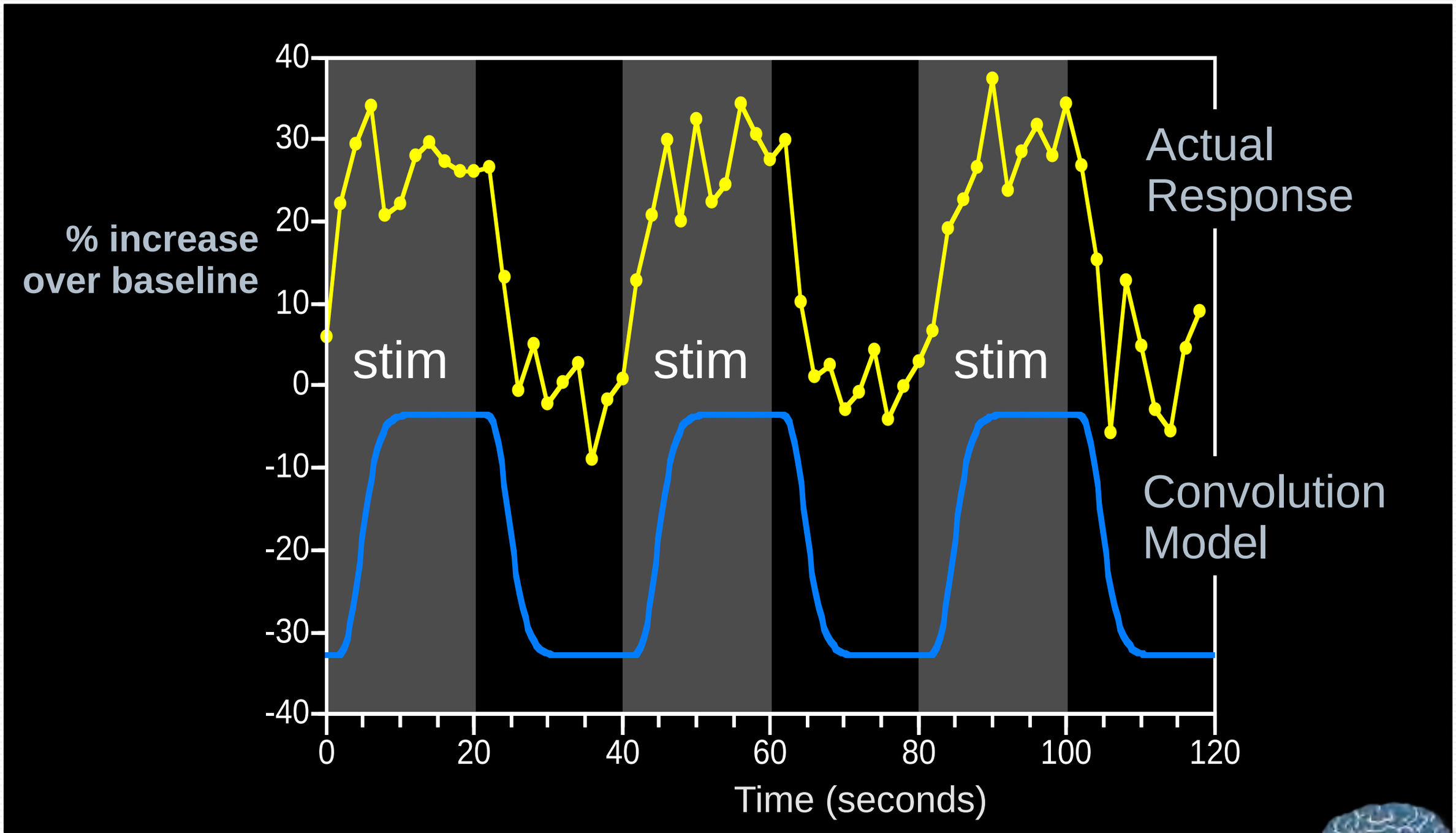


Data courtesy of Robert Savoy

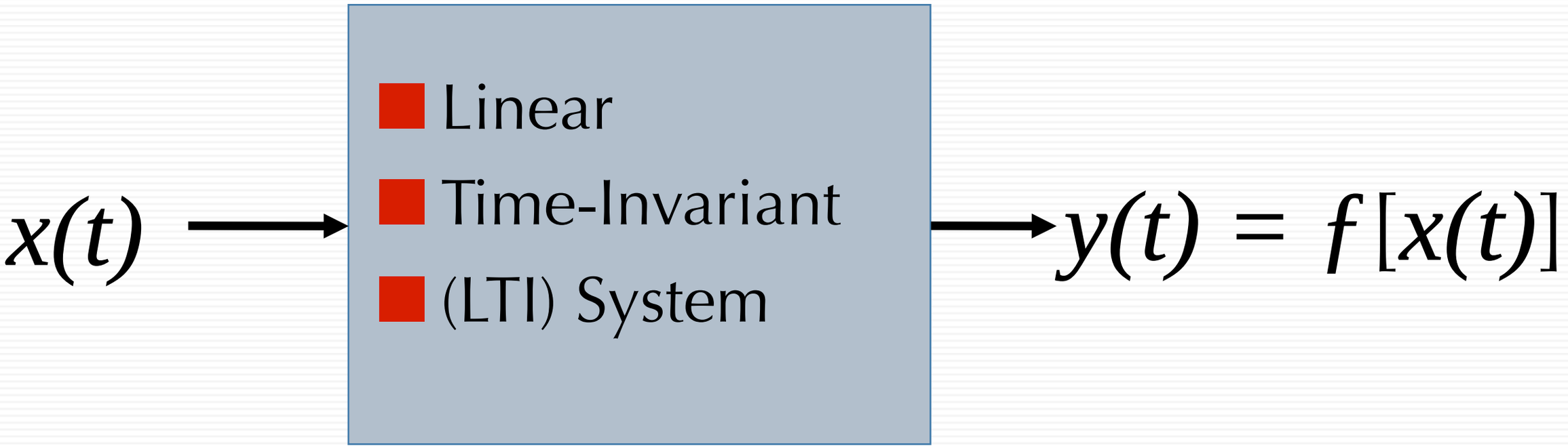
Brain Impulse Response



Convolution of Impulse Responses with Stimuli



Linear Systems Approach



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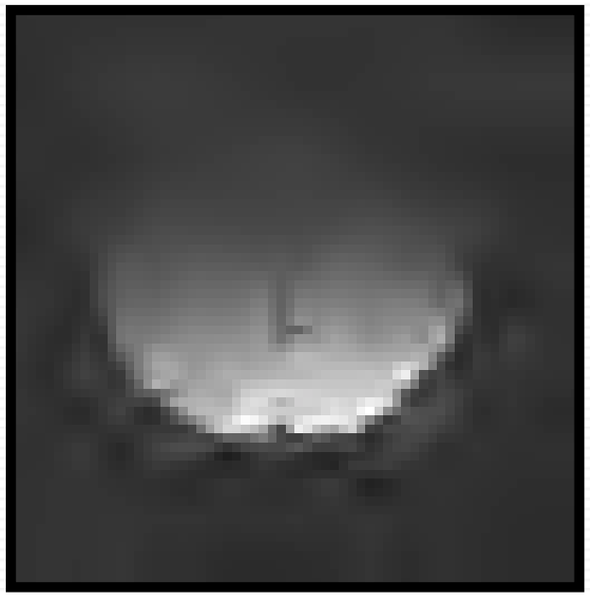
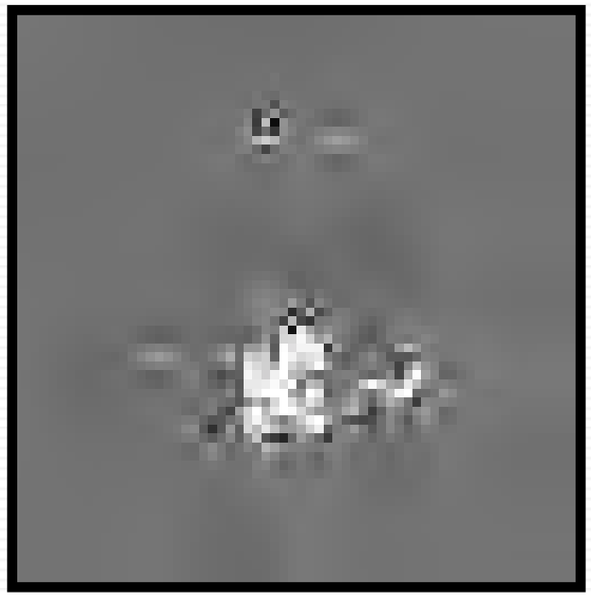
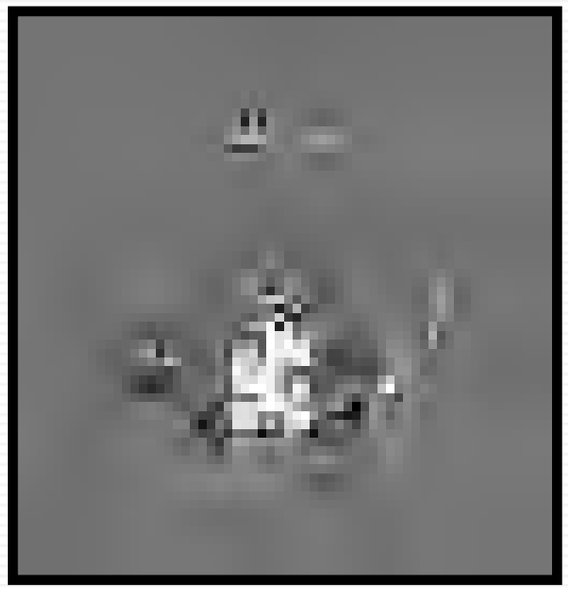
Binocular vs Monocular Activation



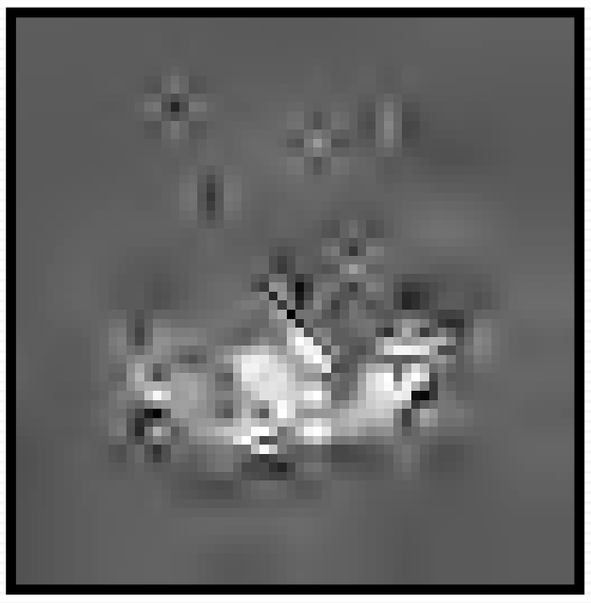
baseline



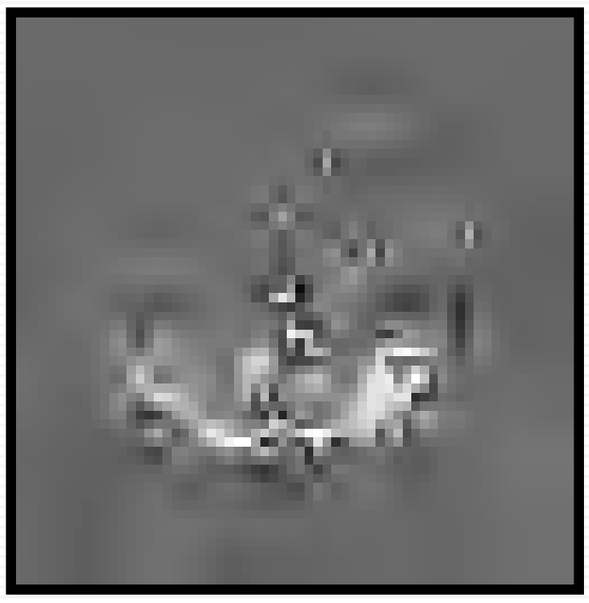
Binocular



Monocular

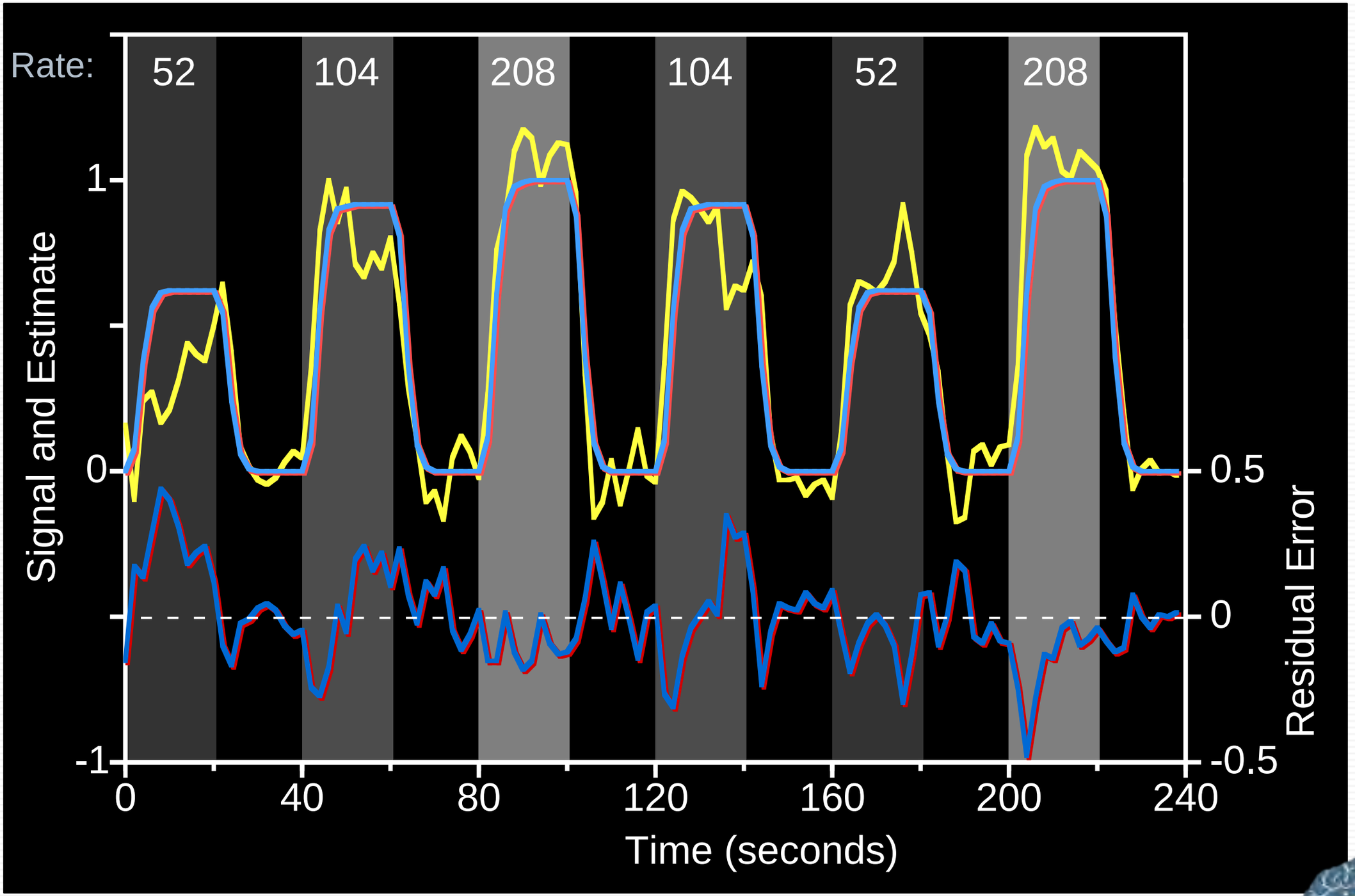


Bino minus Mono

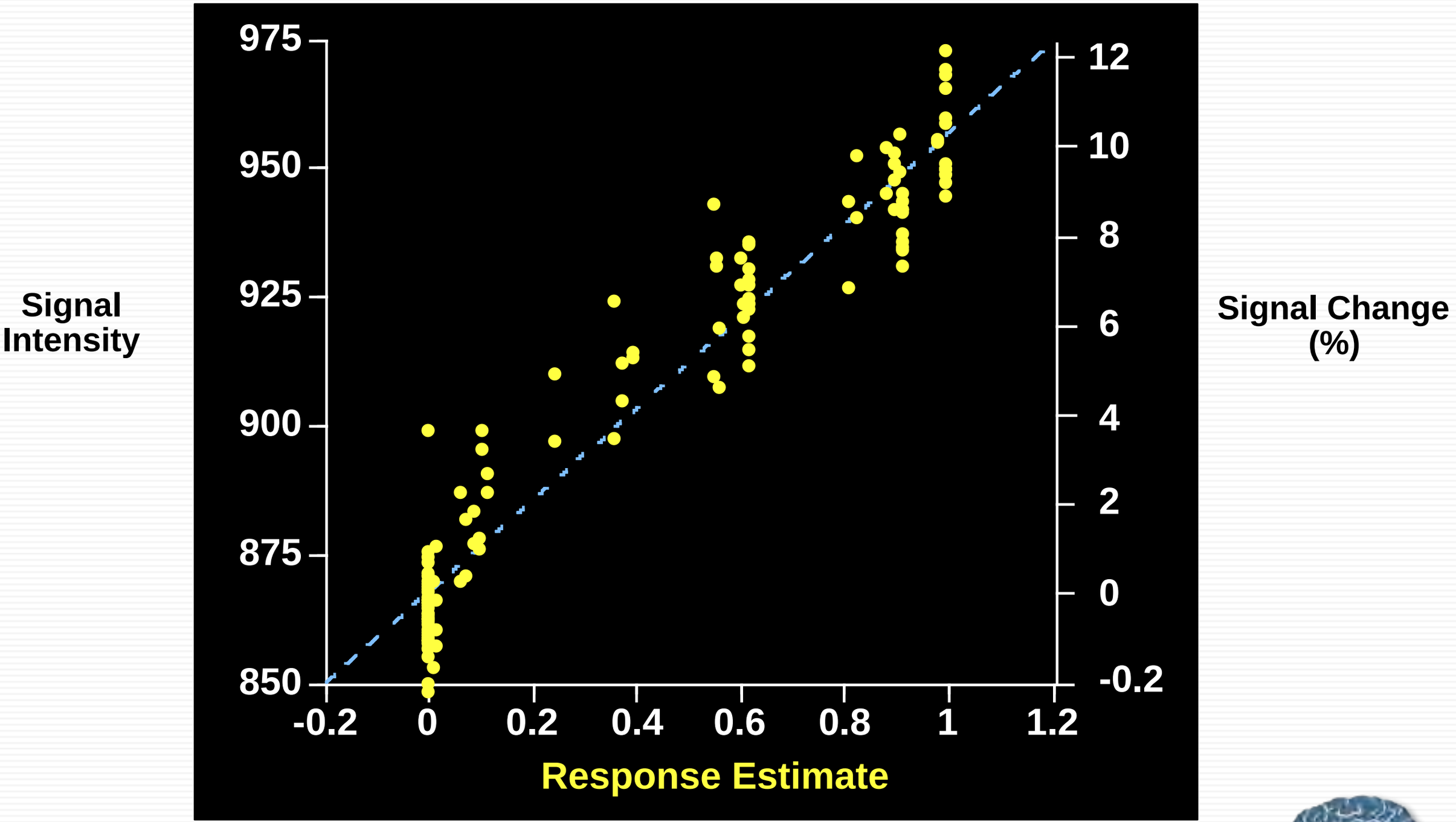


Extrastriate activation

Amplitude-weighted Linear Estimate



Estimated vs. Actual *f*MRI Response



The Plan

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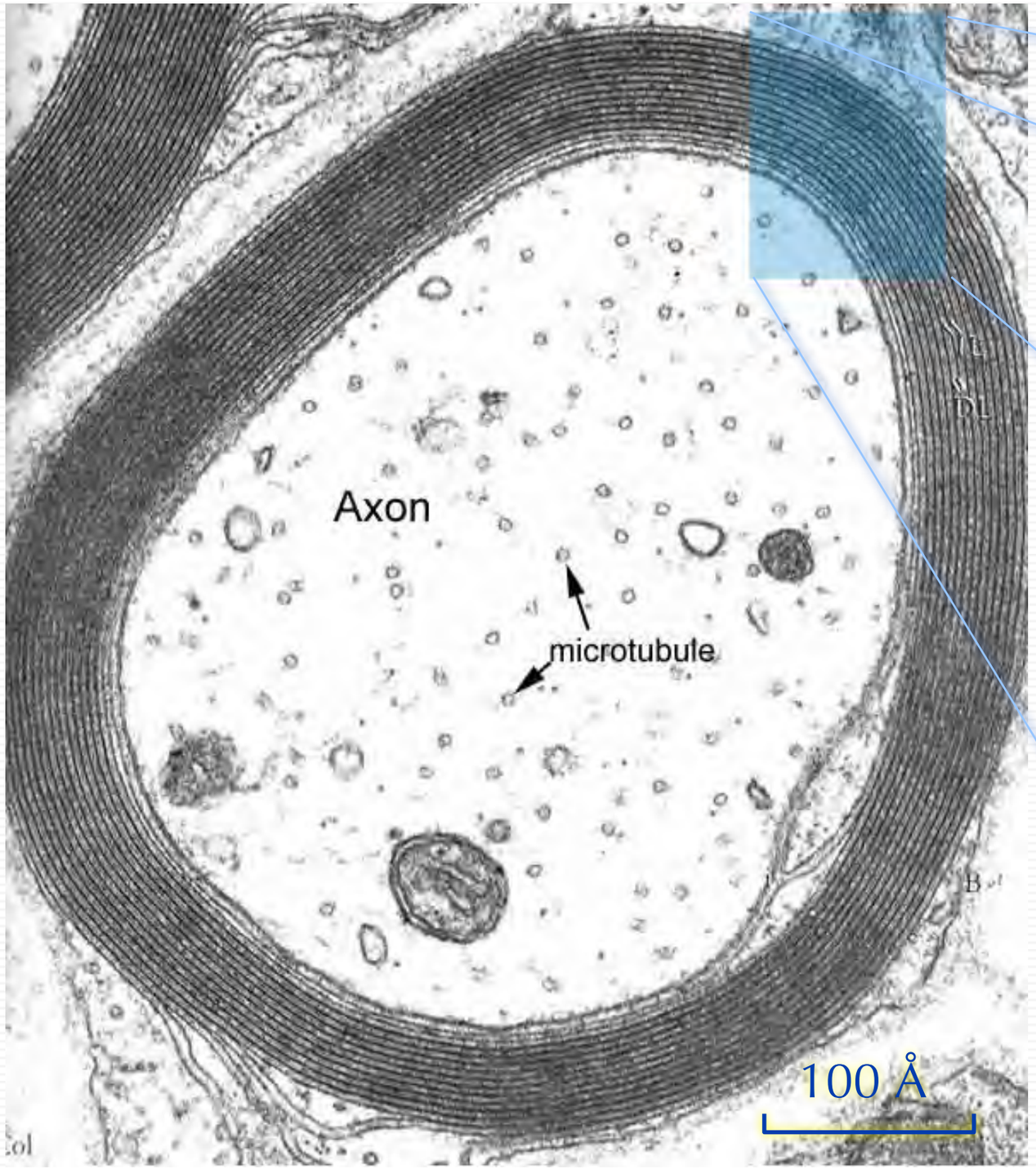
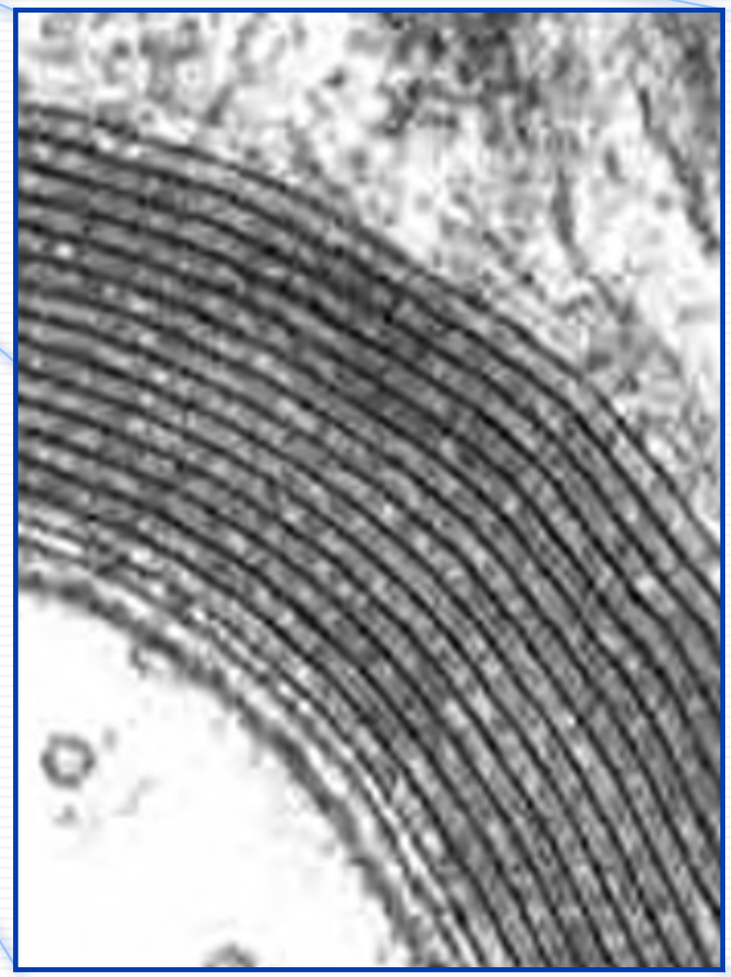
Seventh Inning Stretch

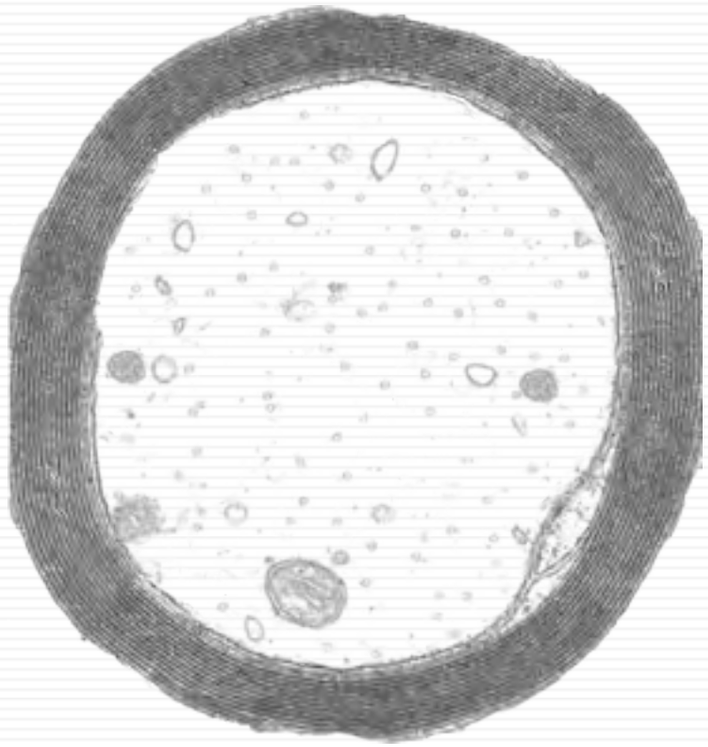
- Fast Imaging (14)
- Functional MRI (18)
- Diffusion and Summary (9)
- Image Quality and Artifacts (48)

Myelin Sheath





Isotropic vs. Anisotropic Diffusion



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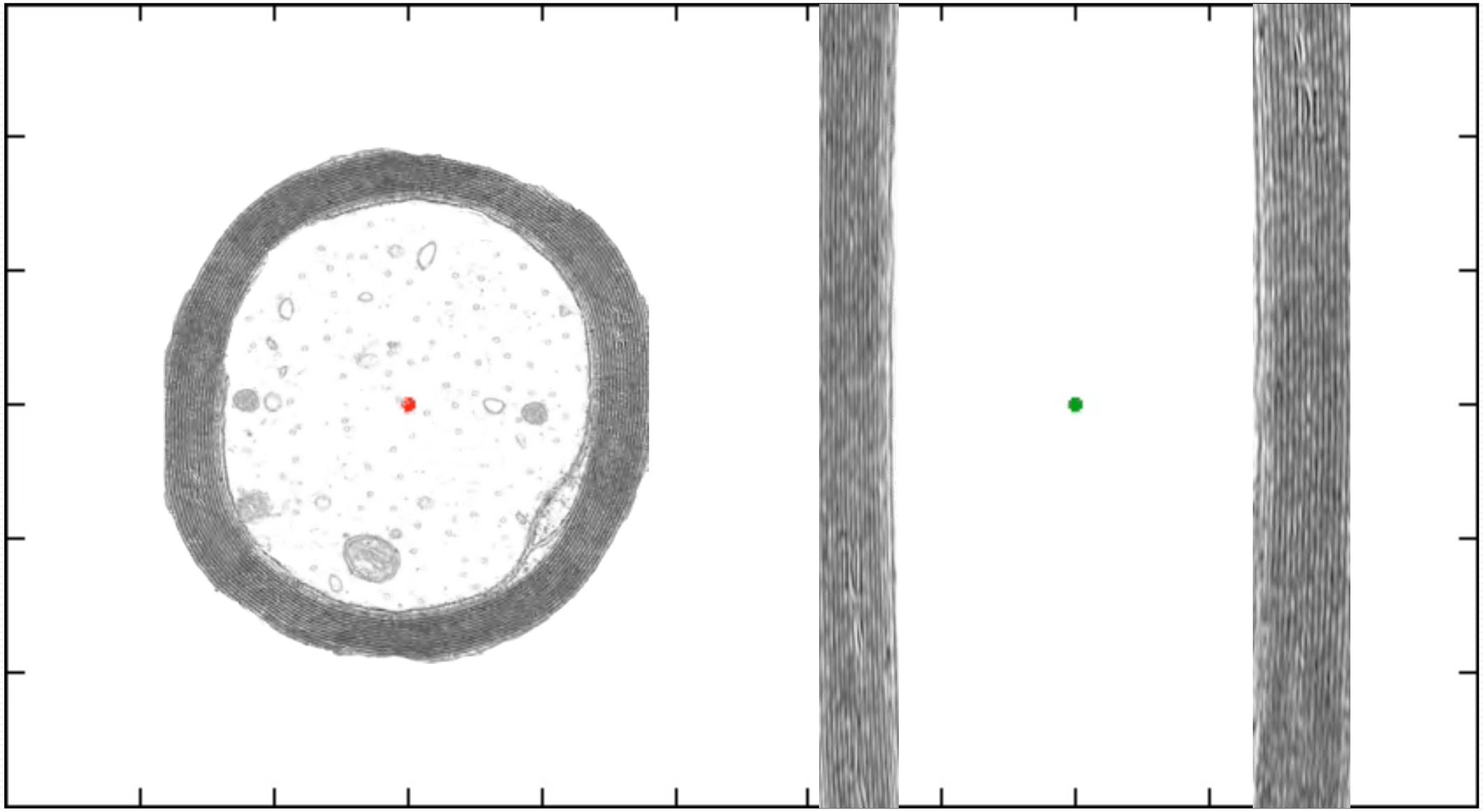
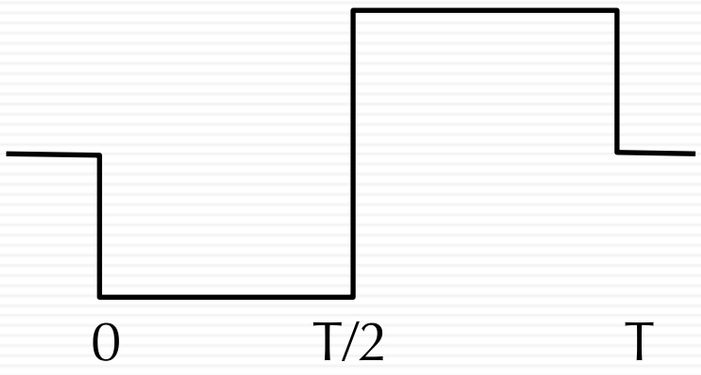


Motion Effects on Phase

- Relative Phase is the product γBt (cycles/sec)/Tesla * Tesla * sec
- Motion causes additional phase shifts:

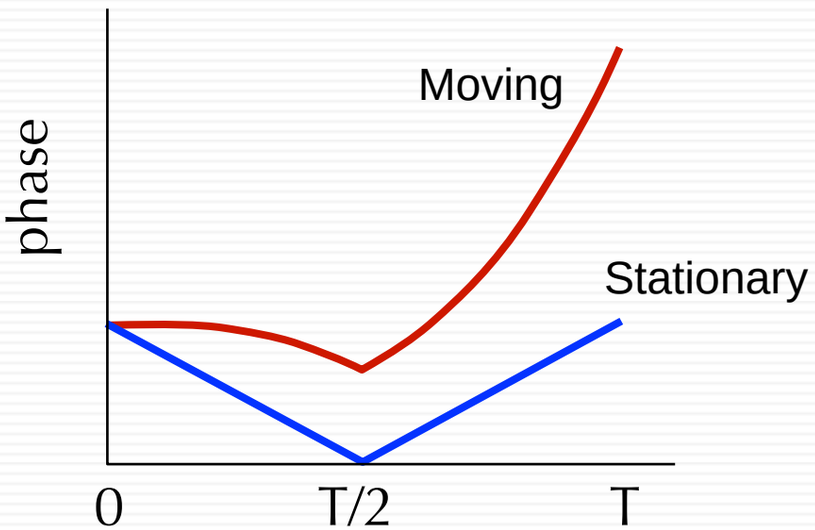


$$x(t) = x_0 + vt + {at^2}/{2} + \ldots$$

$$\varphi = \gamma \int_0^T G(t)x(t)dt$$

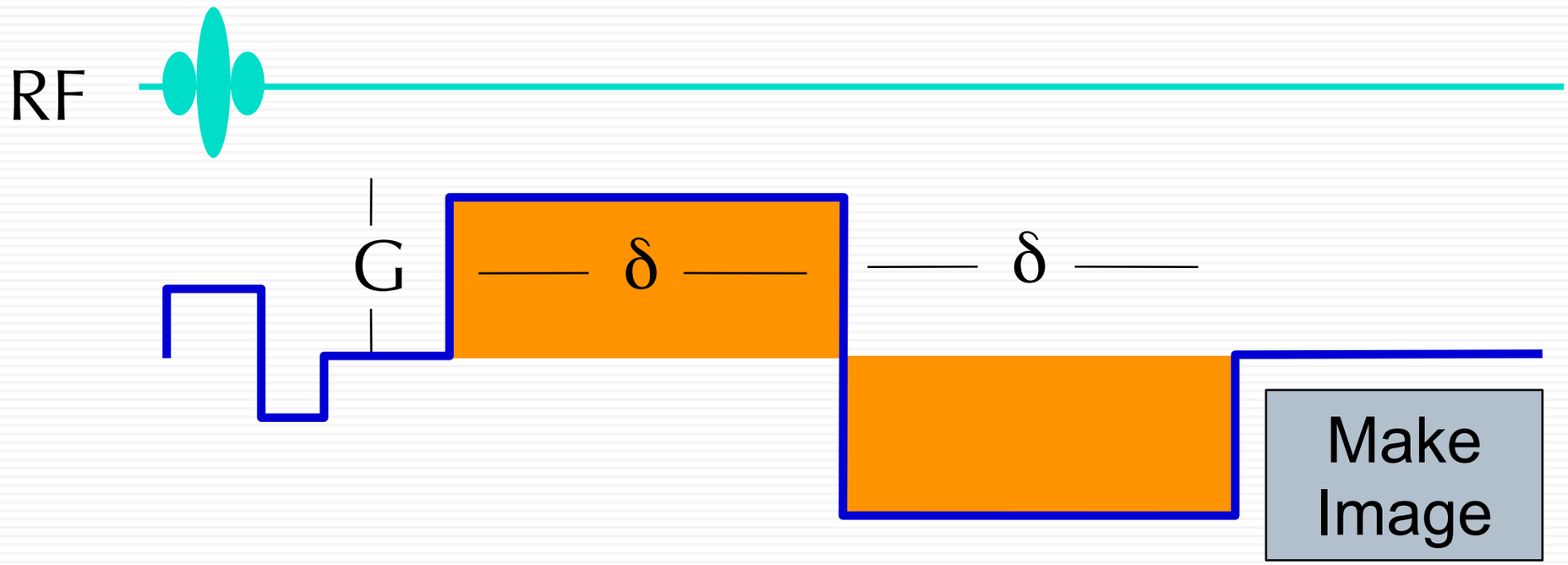
$$= \gamma \int_0^T G(t)[x_0 + vt + {at^2}/{2} + \ldots]dt$$

$$= \gamma \left[{vT^2}/{4} + 23{aT^3}/{24} + \ldots \right]$$



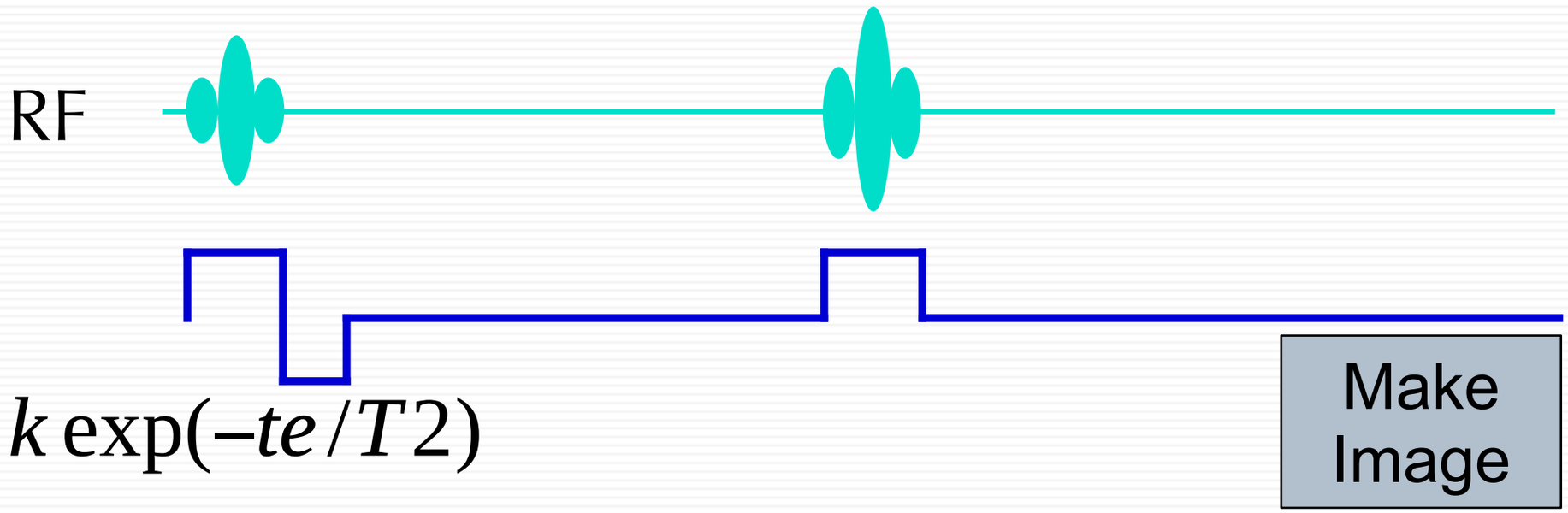
Diffusion Gradients and Signal

Make Image



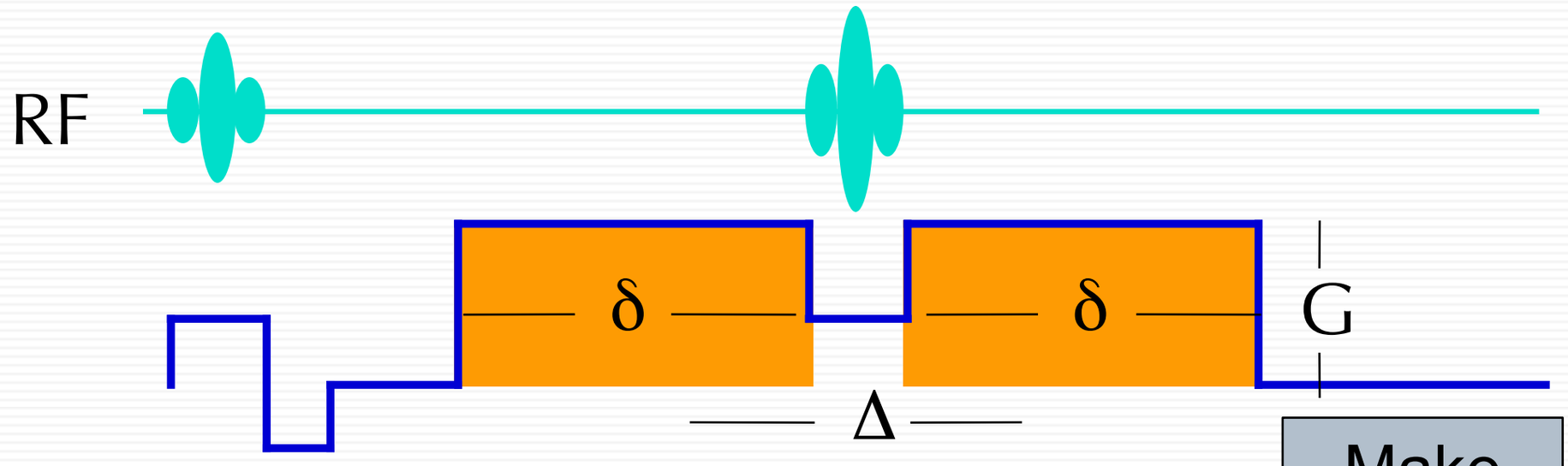
Make Image

Diffusion Gradients and Signal



$$SI_0 = k\exp(-te/T2)$$

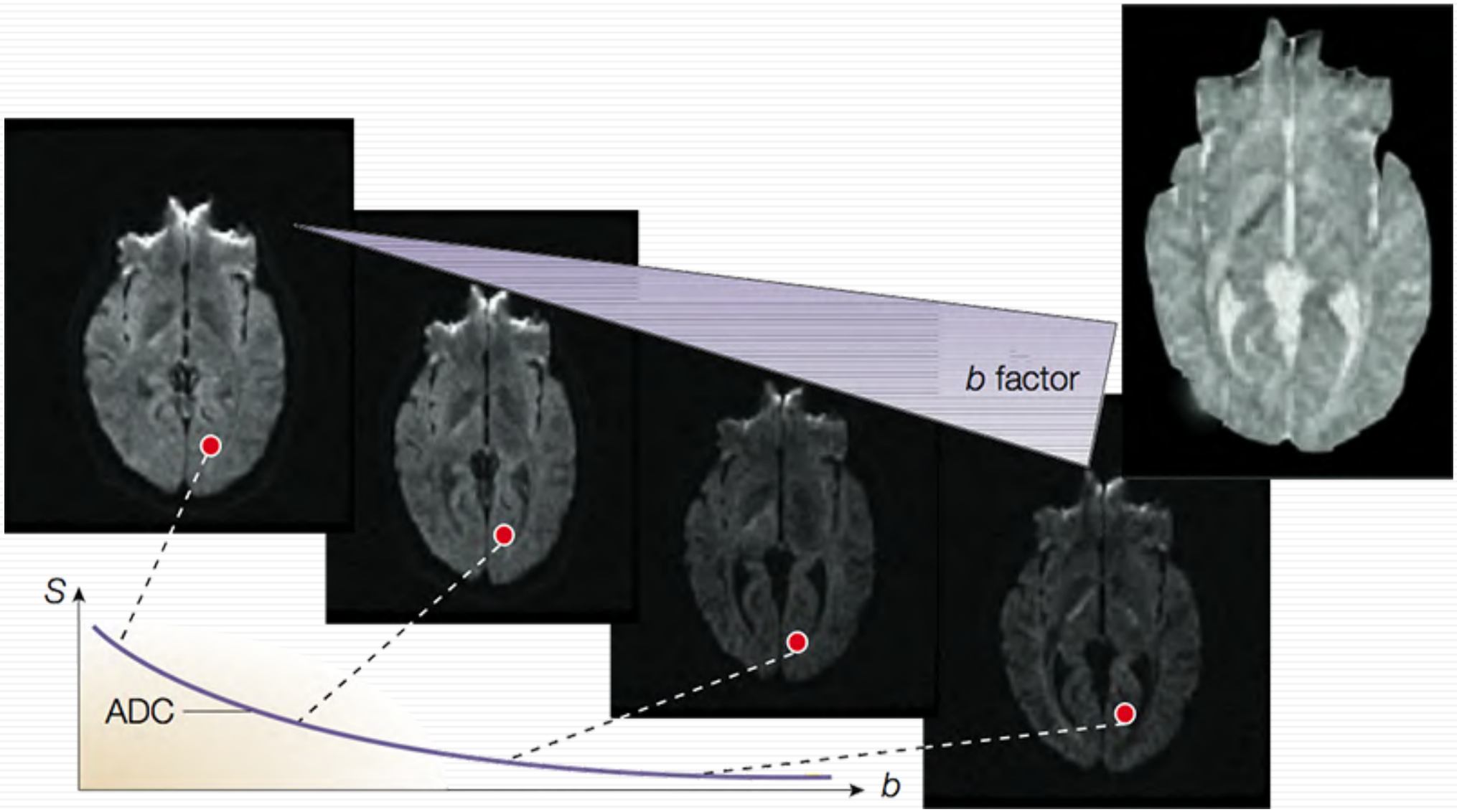
Make Image



$$SI_D = k\exp(-te/T2)\cdot\exp[-\gamma^2 G^2\delta^2(\Delta-\delta)]$$

Make Image

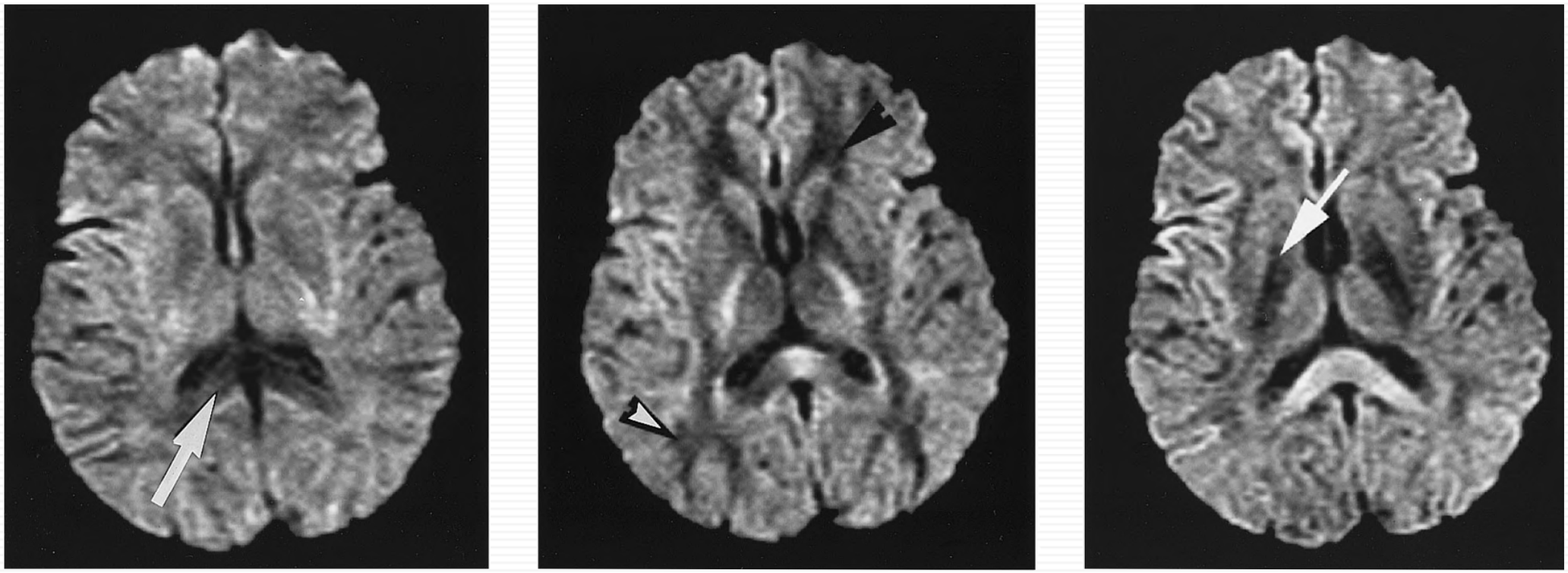
Diffusion Attenuates MR Signal



Denis Le Bihan, Nature Reviews in Neuroscience 4:469, 2003

Brain Diffusion is Anisotropic

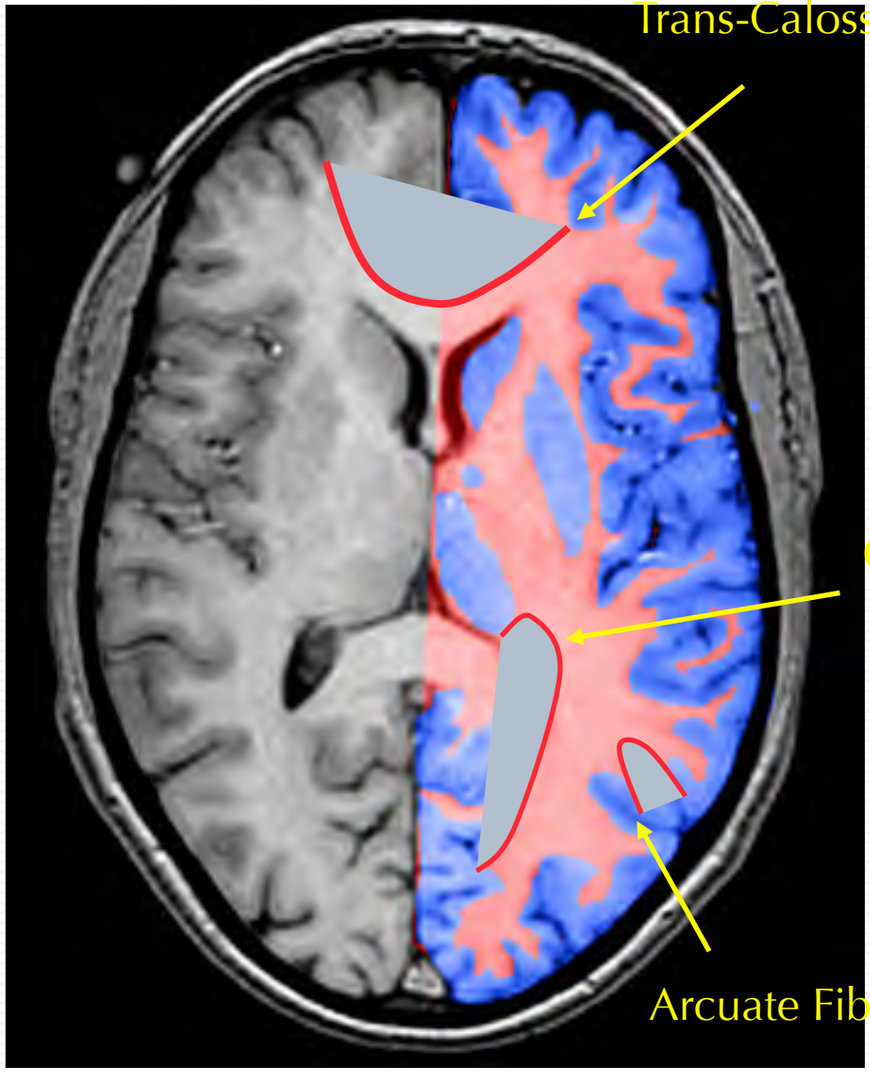
Radiology

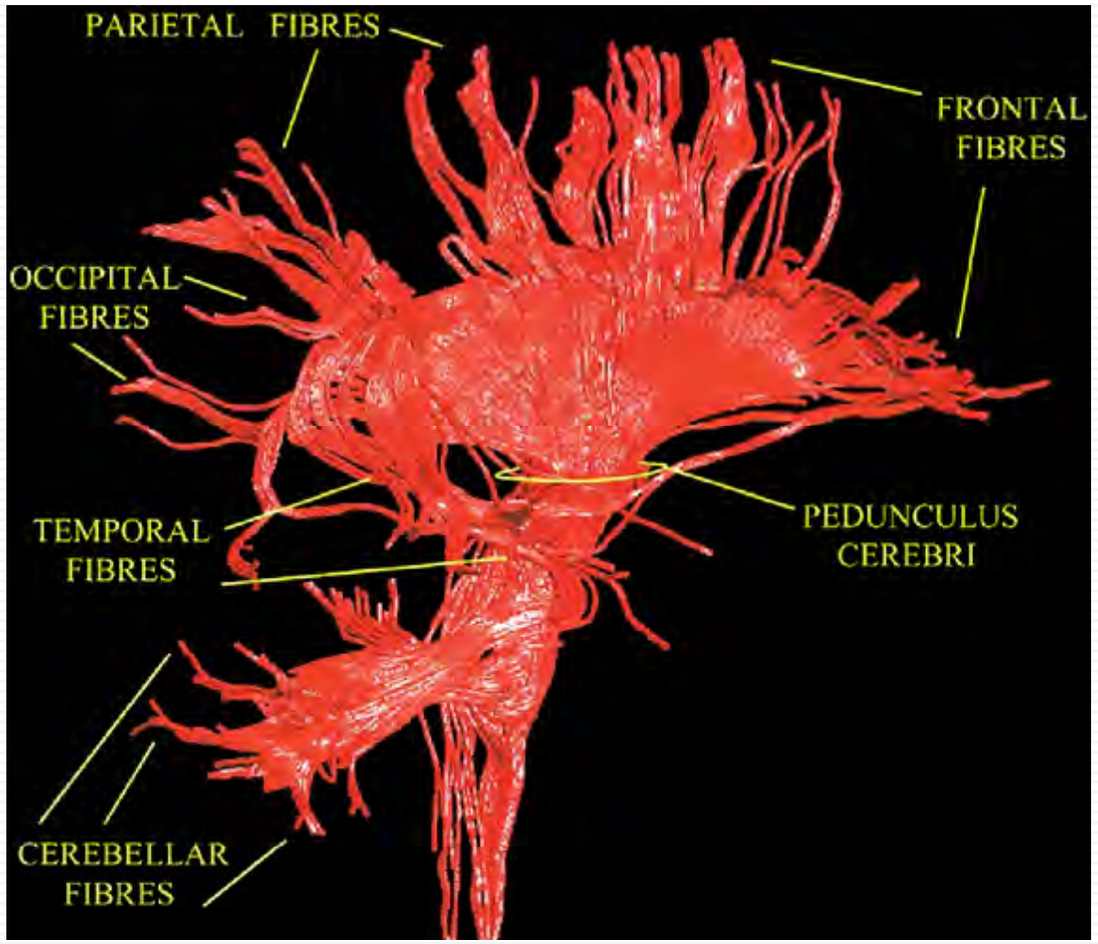


Gx Gy Gz

Schaefer, P. W. et al. Radiology 2000;217:331-345

White and Gray Matter



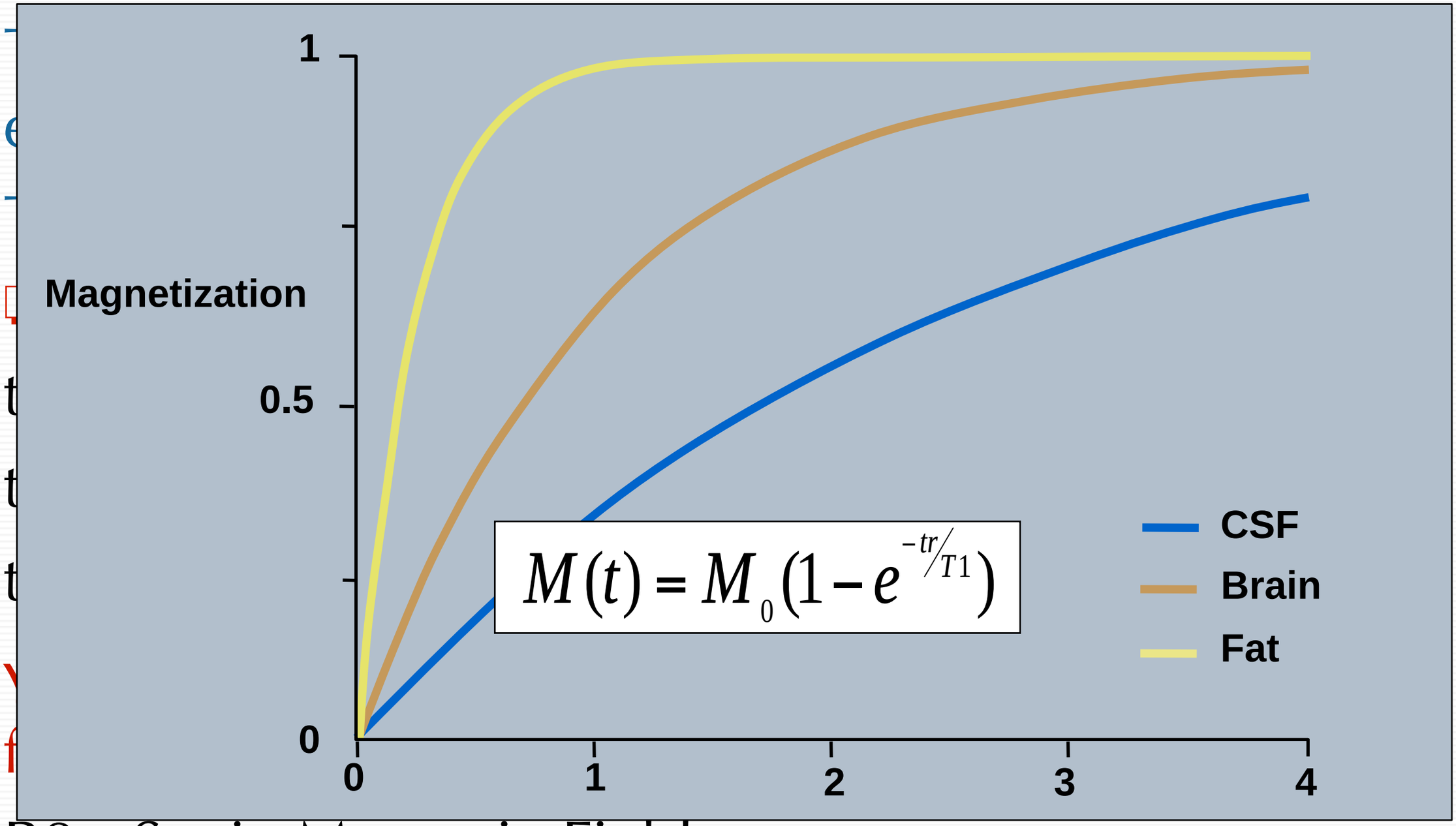


After: Catani, et al., NeuroImage 17:77, 2002

Crib Sheet

- T1 - Longitudinal Relaxation Rate (reaching equilibrium)
- T2 - Transverse Relaxation Rate (dephasing)
  - T2* - observed T2 decay, made up of T2, T2′, T2D, etc…
- tr - Repetition time between excitation pulses
- te - Time after excitation before forming images
- ti - Time between inversion and excitation
- γ - Gyromagnetic ratio - proportionality of field and frequency
- B0 - Static Magnetic Field
- B1 - Rotating Magnetic Field

Crib Sheet



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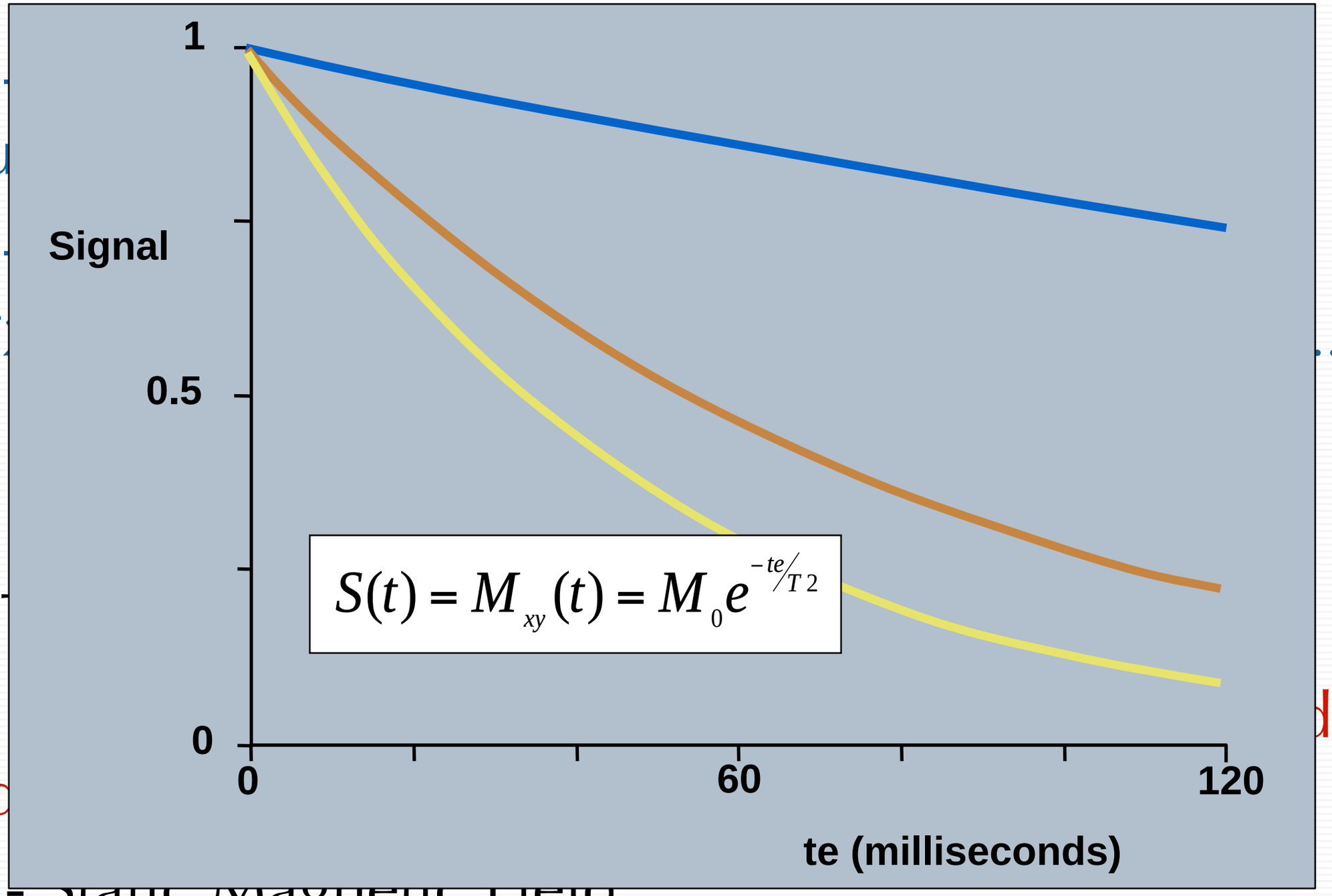

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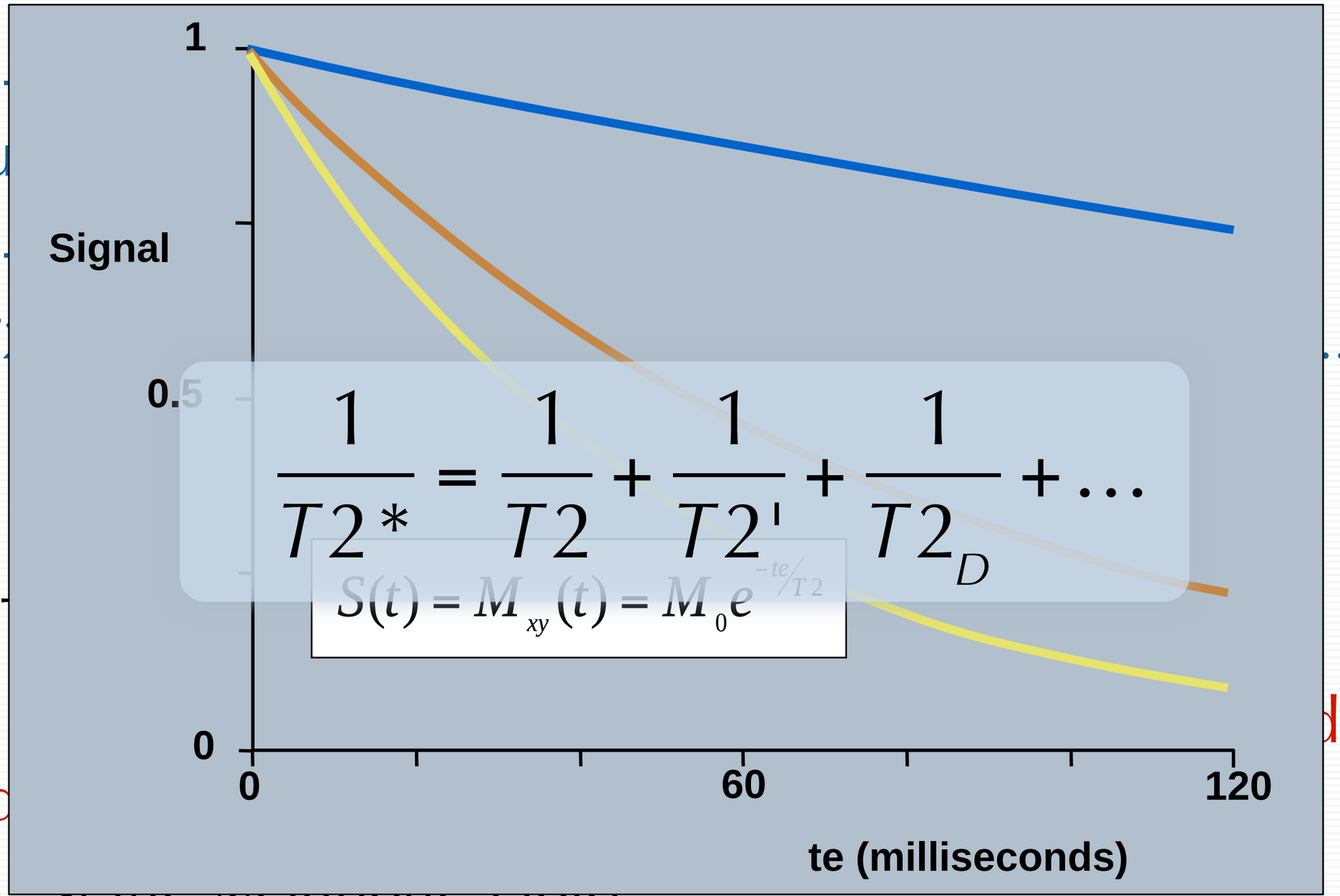

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  - T2*
- tr - Re
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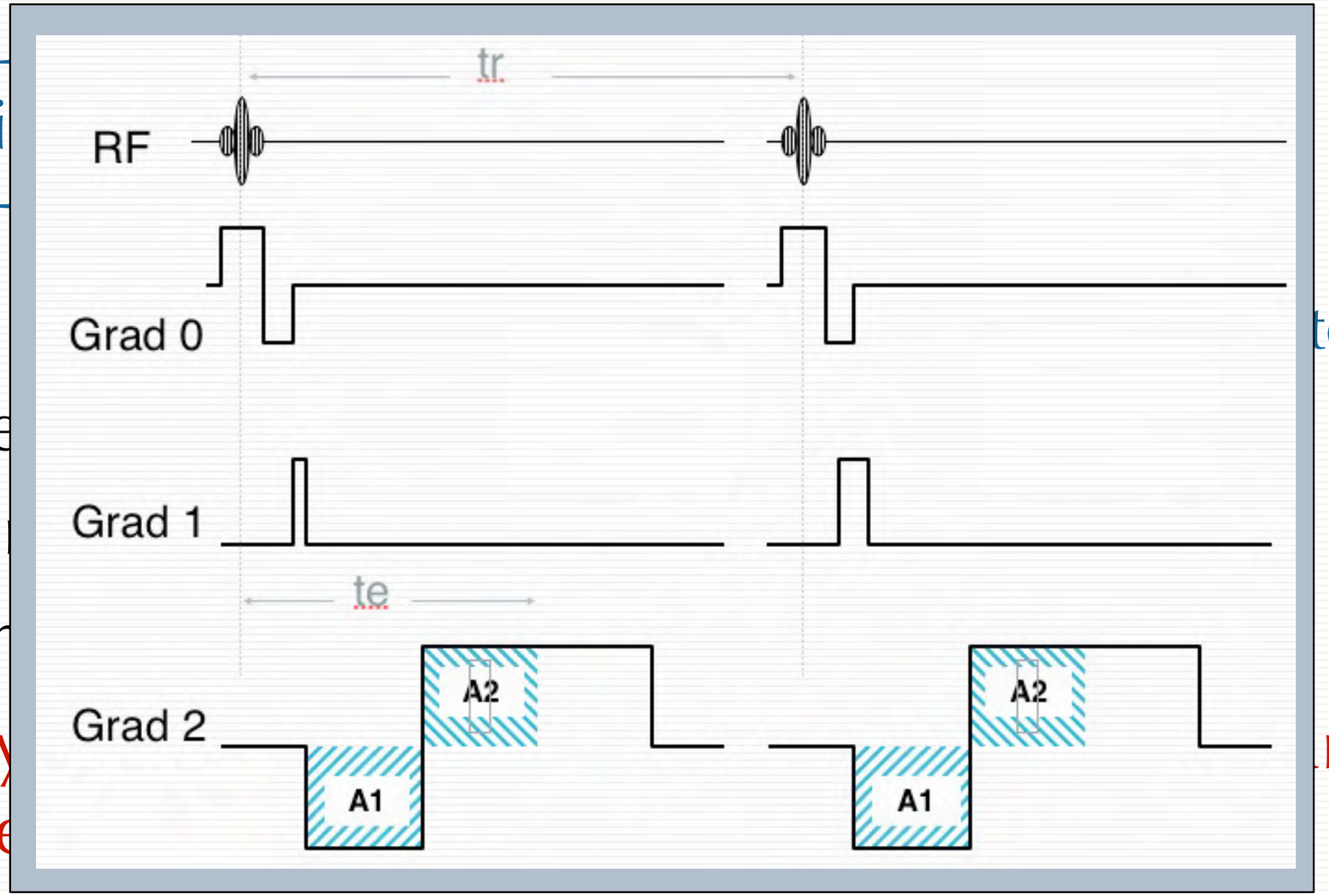

tc…

nd

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- T1
  - equ
- T2
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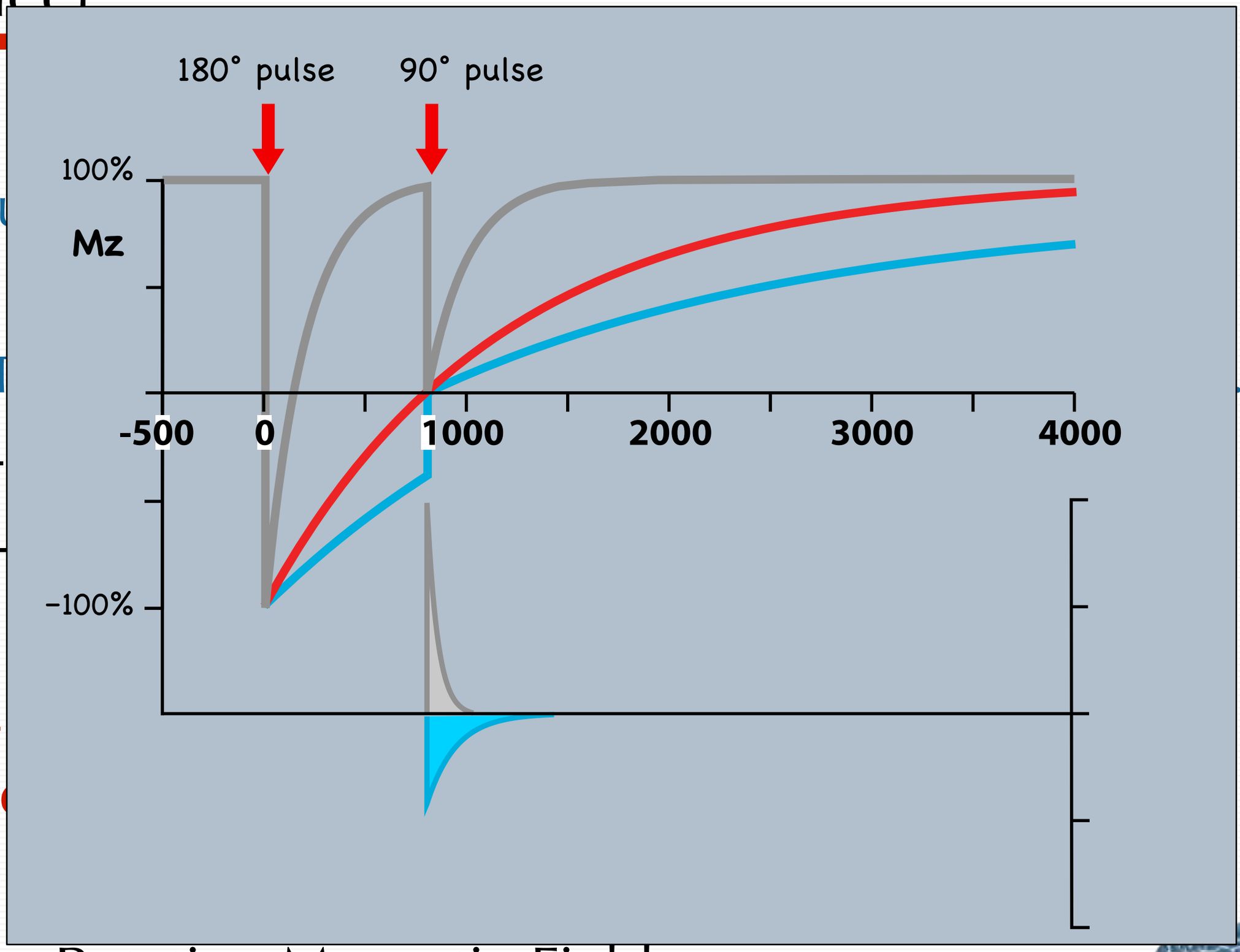
- tr -
- te -
- ti -
- γ -
  - free
- B0
- B1 - Rotating Magnetic Field



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- T
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- tr
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- B
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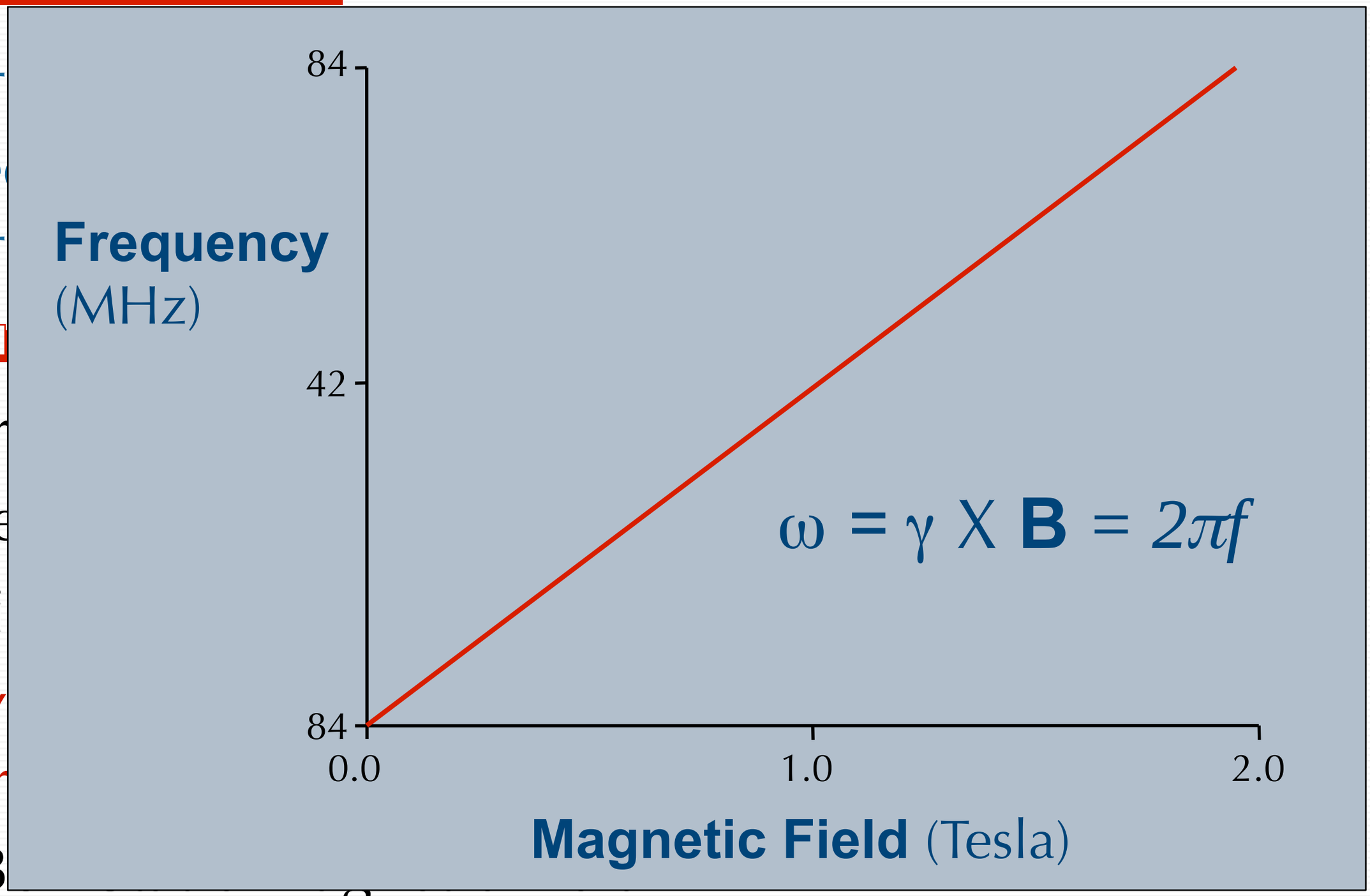

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  - T2* -
- tr - Rep
- te - Tim
- ti - Tim
- γ - Gyr
  frequenc
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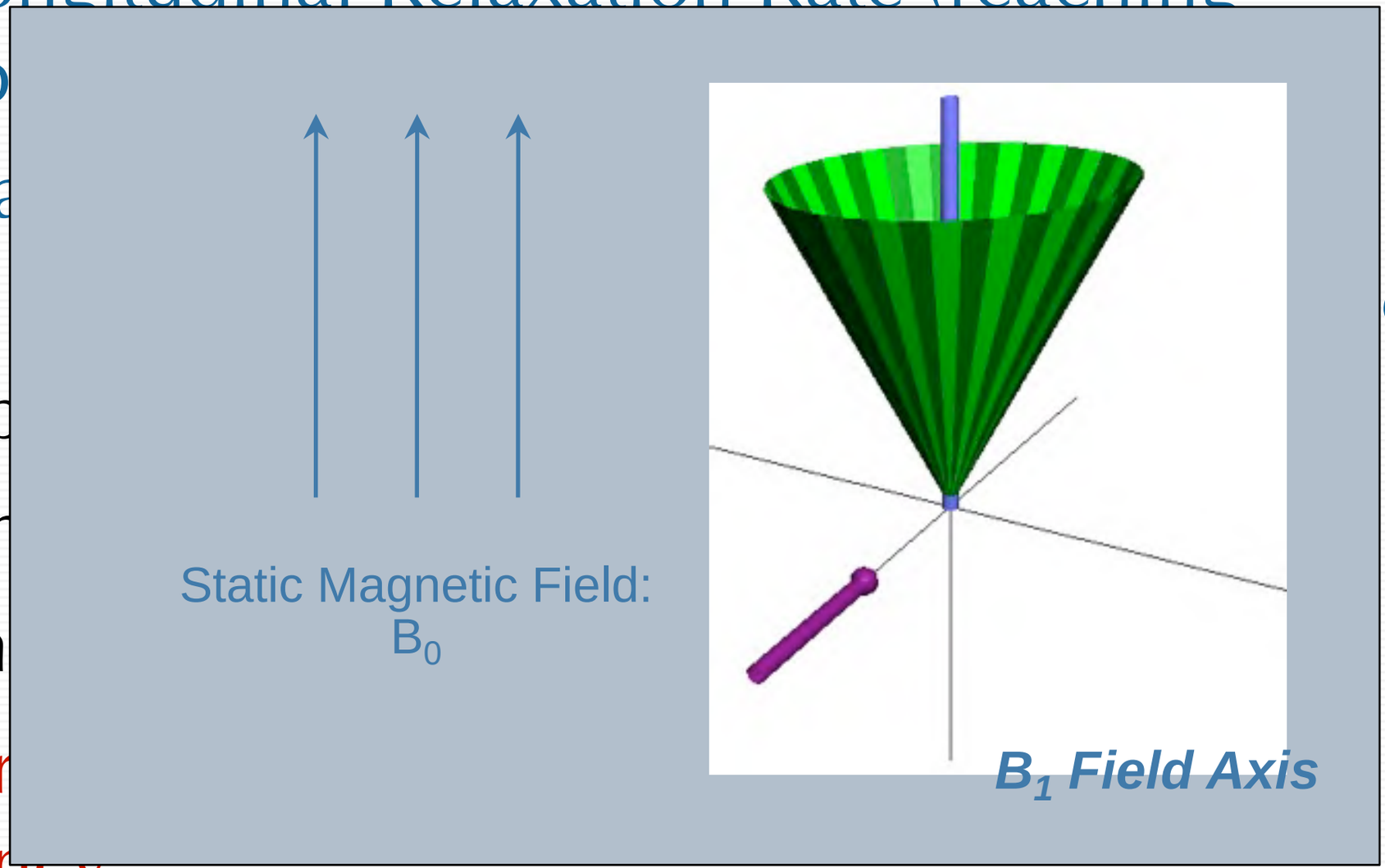
etc…

and



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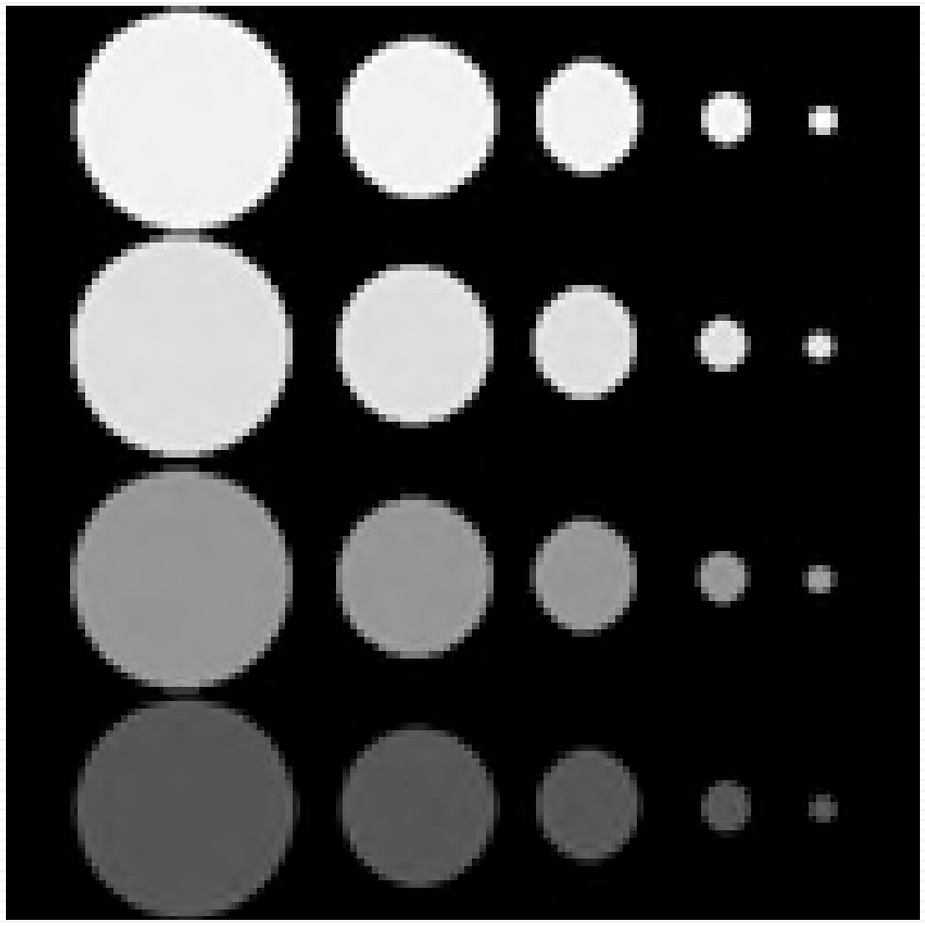
An "Equation" in Resolution

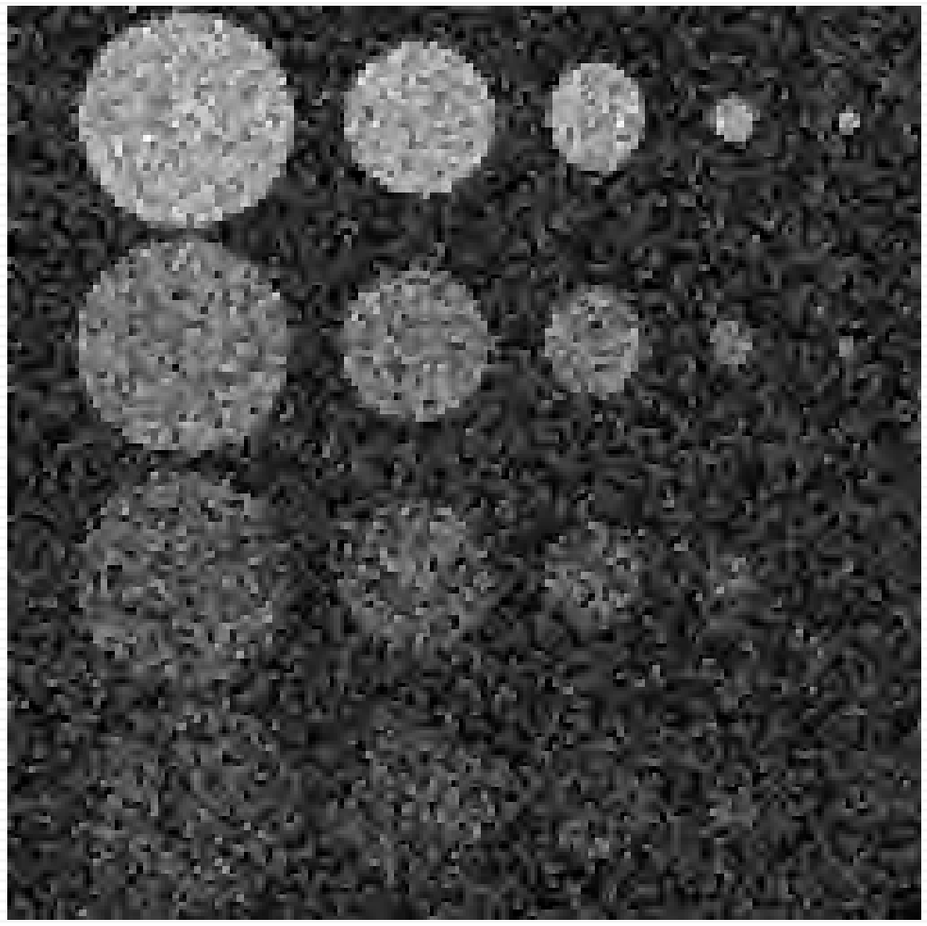
Because MR is an emission modality the temporal resolution, spatial resolution and contrast are inter-dependent:

$$\text{Signal} = kB_0(\text{voxel size})\sqrt{\text{imaging time}}$$
$$-\text{contrast}$$

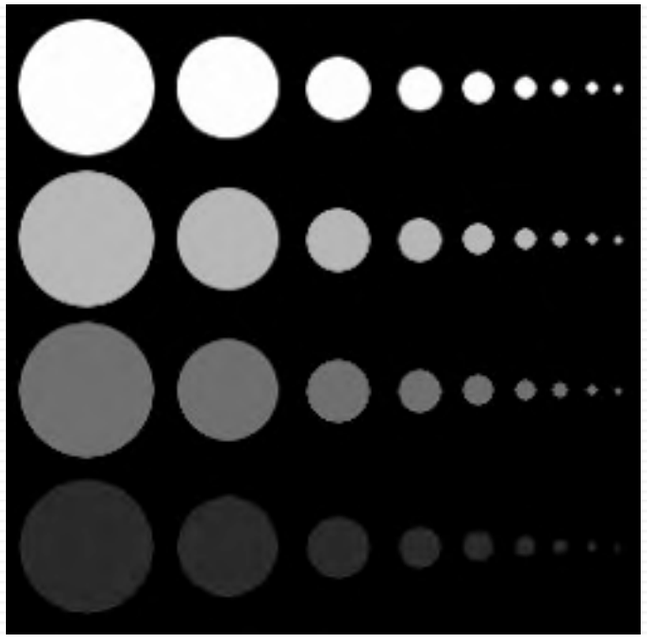
where B_0 is the field strength.

Contrast to Noise Ratio



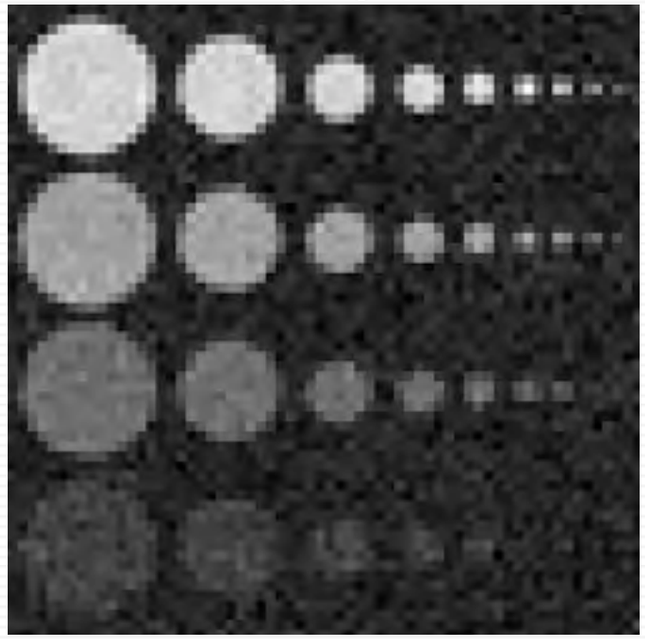


CNR vs. Resolution



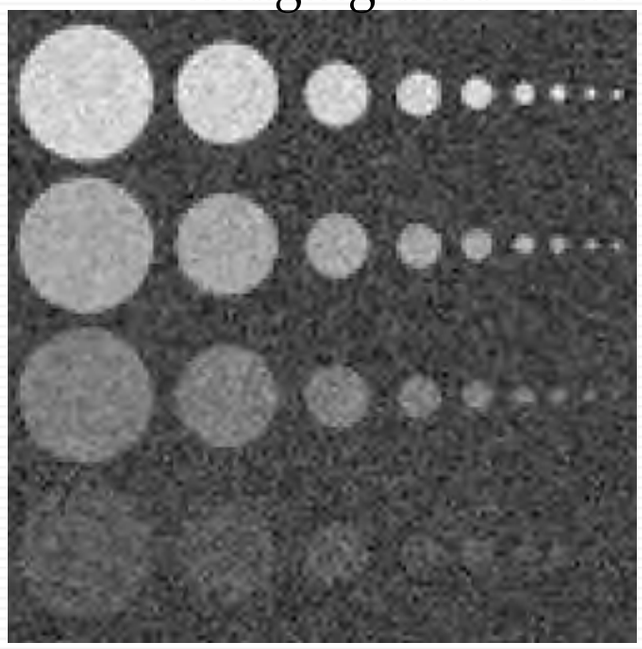
Noise free

Imaging time = 1X



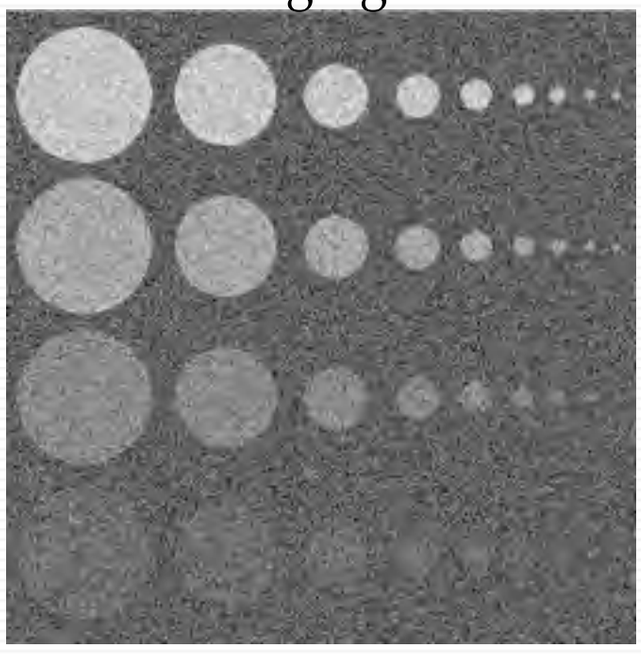
64 X 64

Imaging time = 2X



128 X 128

Imaging time = 4X



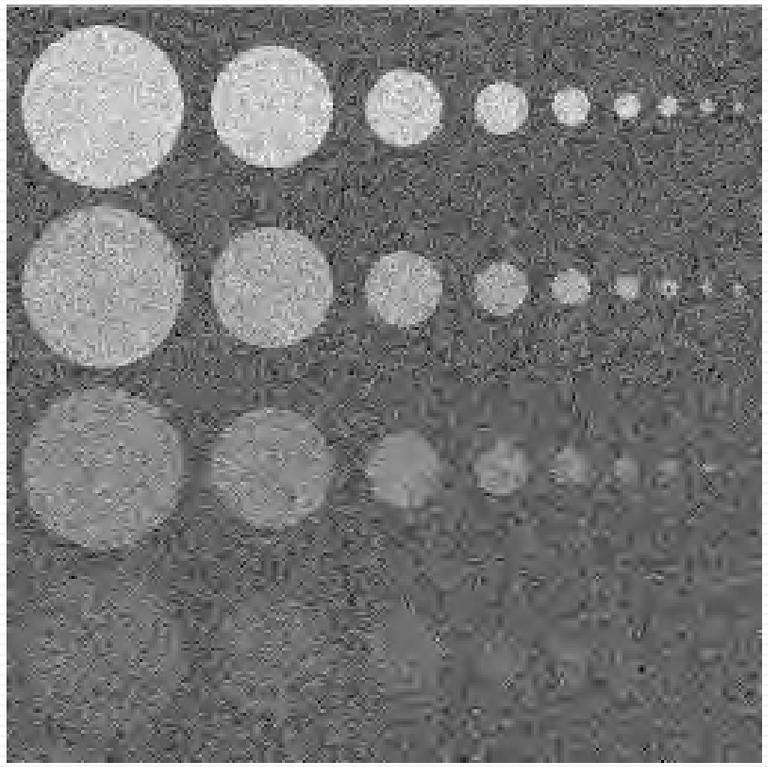
256 X 256

Minimum Imaging Time

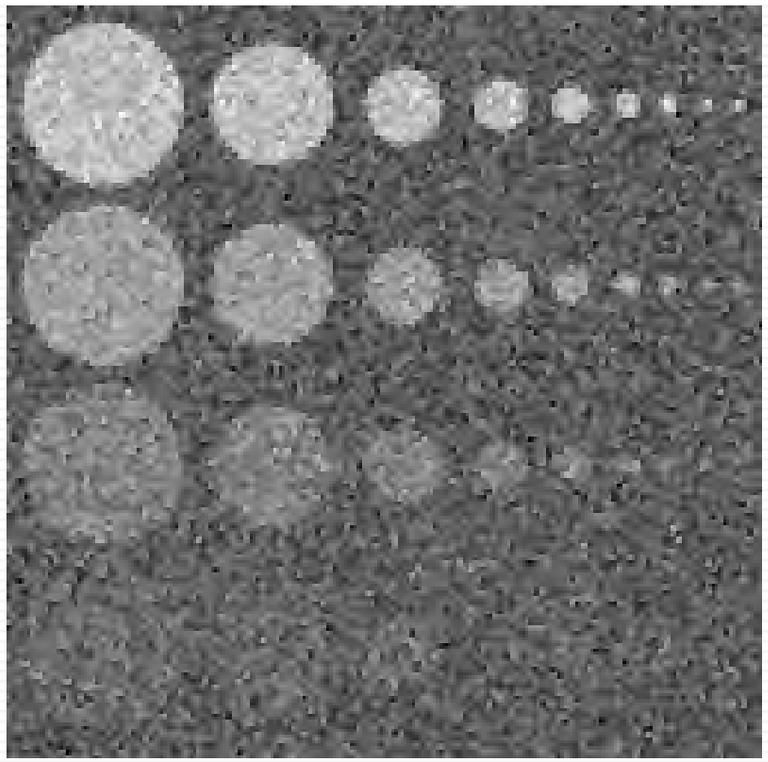
CNR vs. Resolution

Signal/Noise Ratio Held Constant

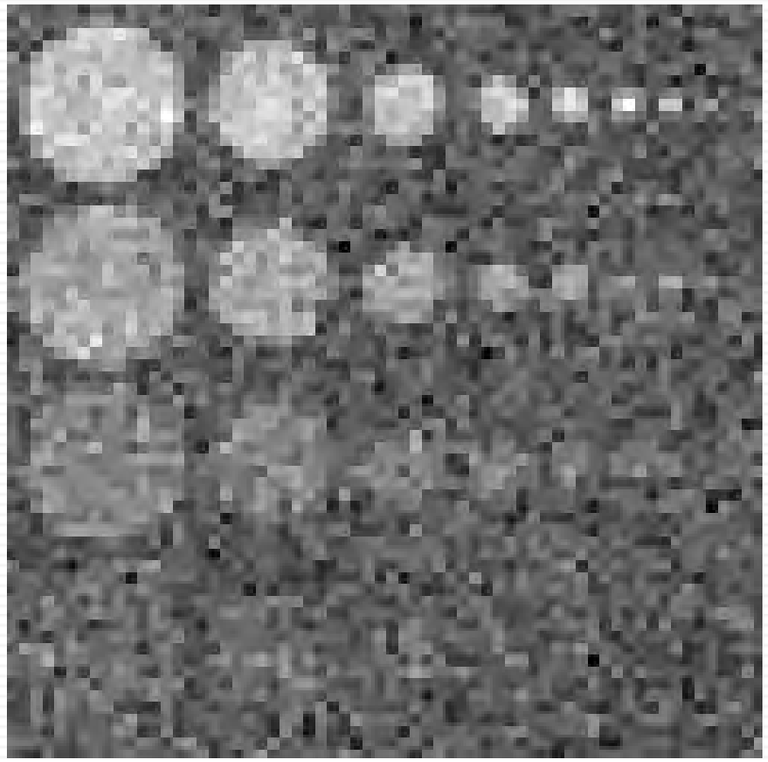
256 X 256



128 X 128



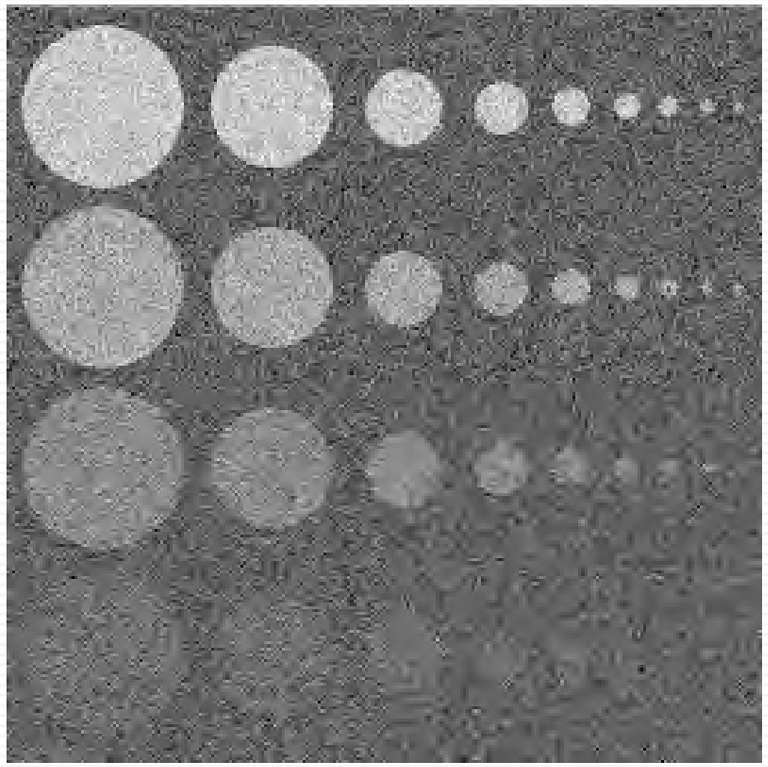
64 X 64



CNR vs. Resolution

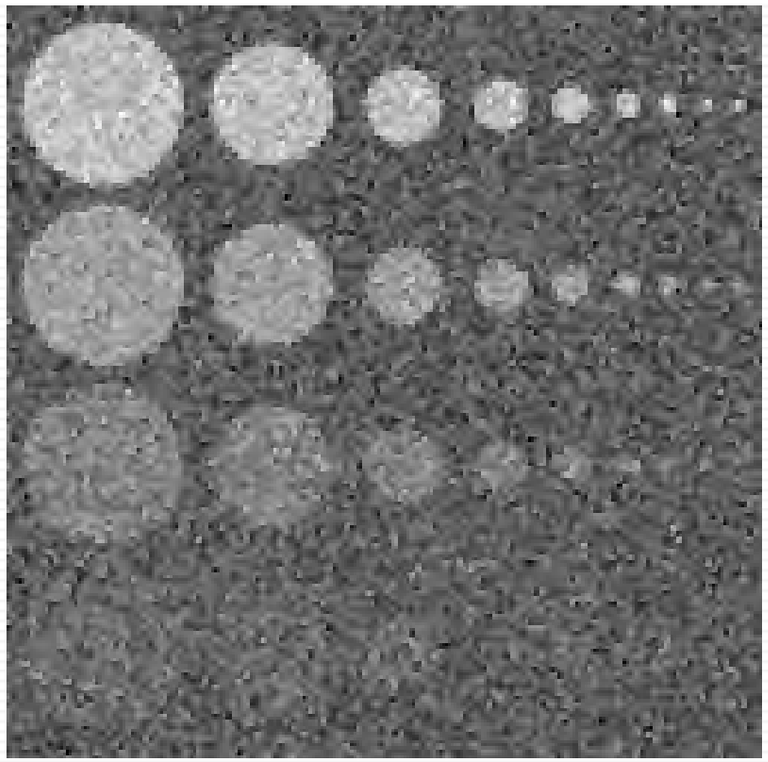
Signal/Noise Ratio Held Constant

256 X 256



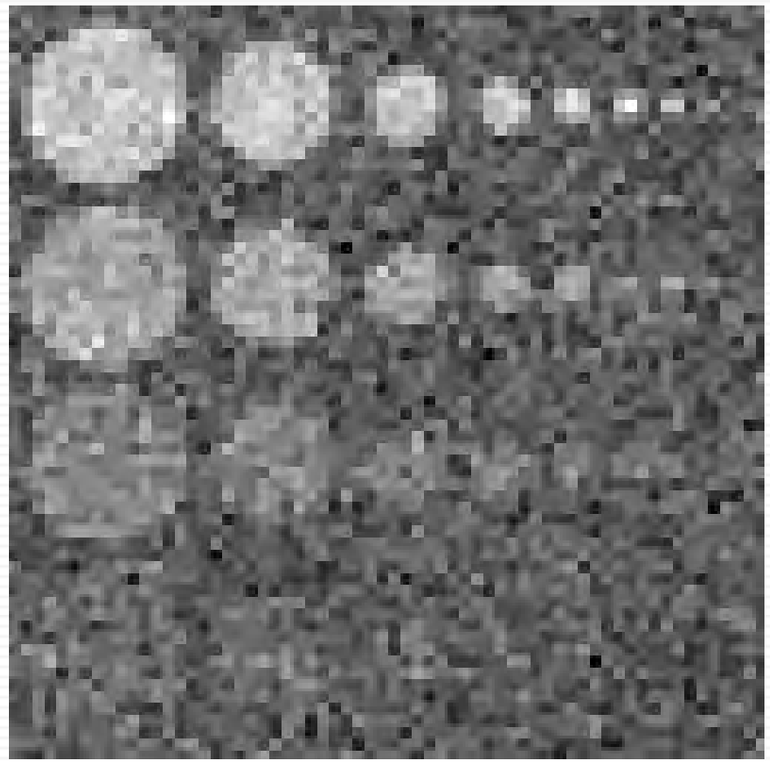
Imaging time = 16X

128 X 128



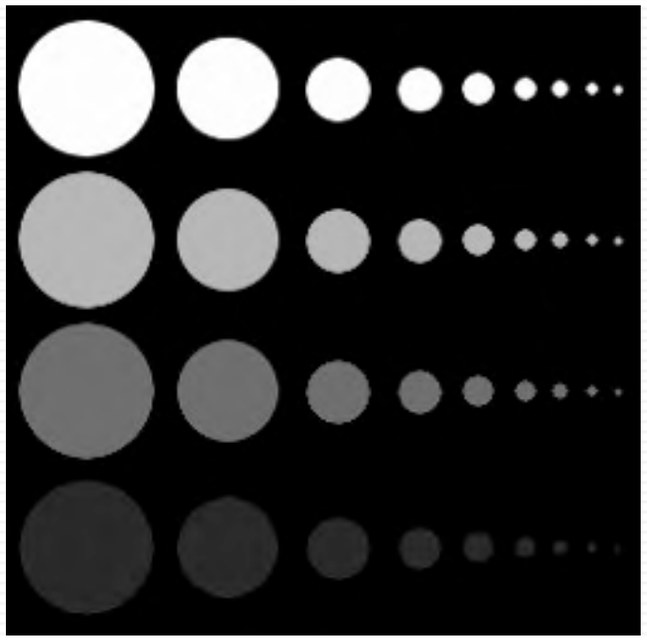
Imaging time = 4X

64 X 64



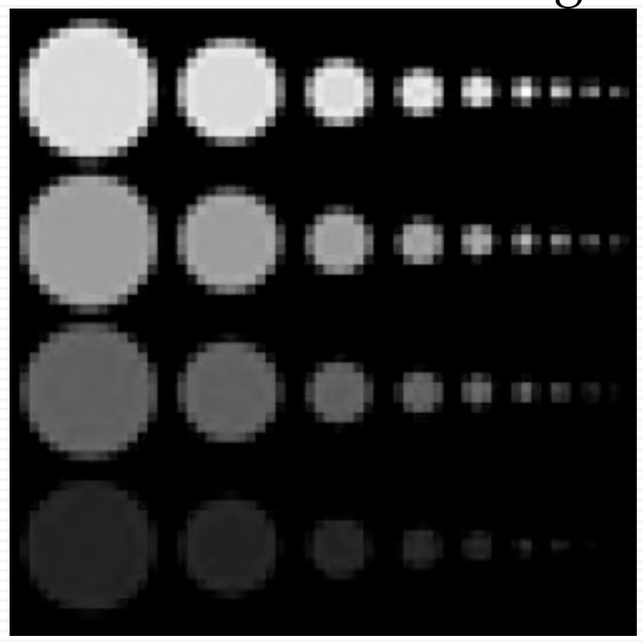
Imaging time = 1X

CNR vs. Resolution



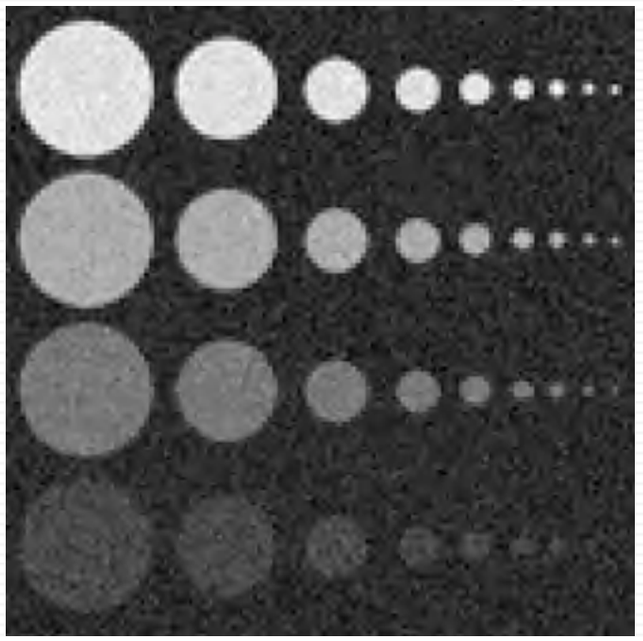
Noise free

16 averages



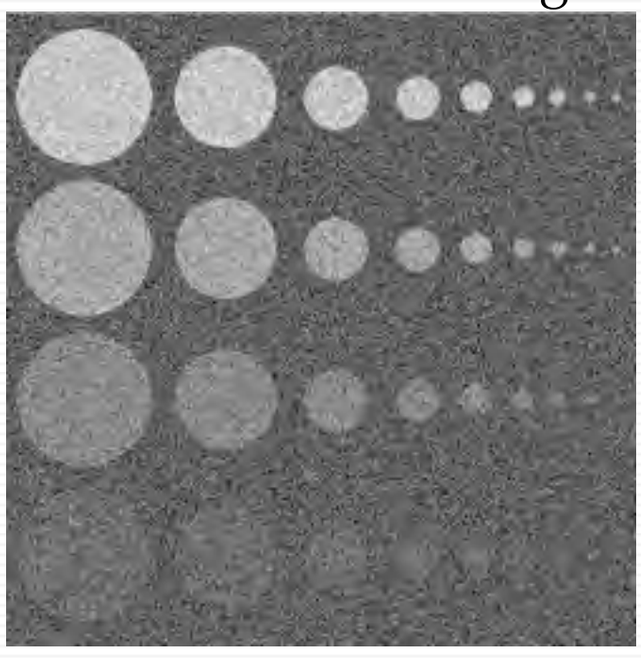
64 X 64

4 averages



128 X 128

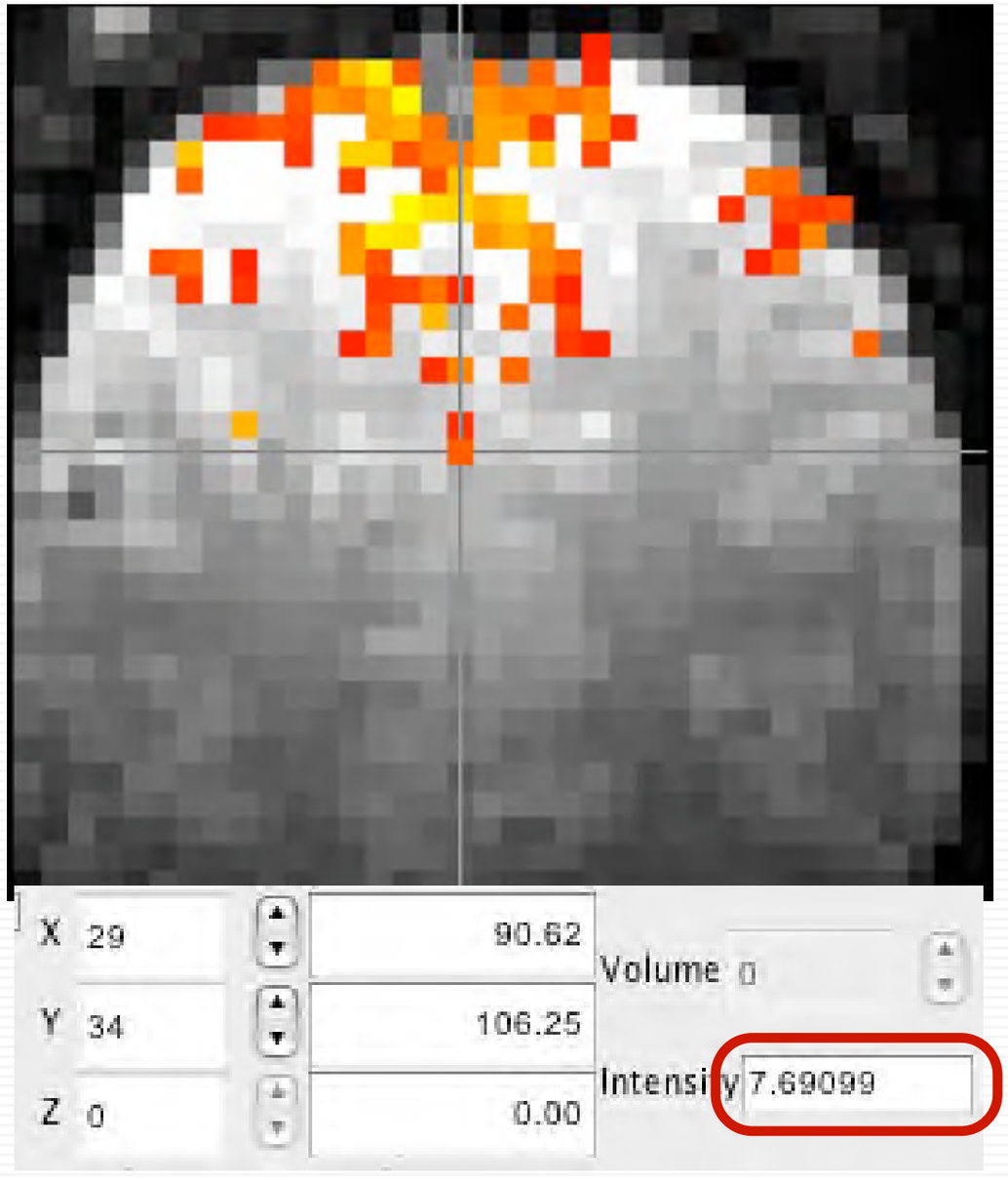
1 average



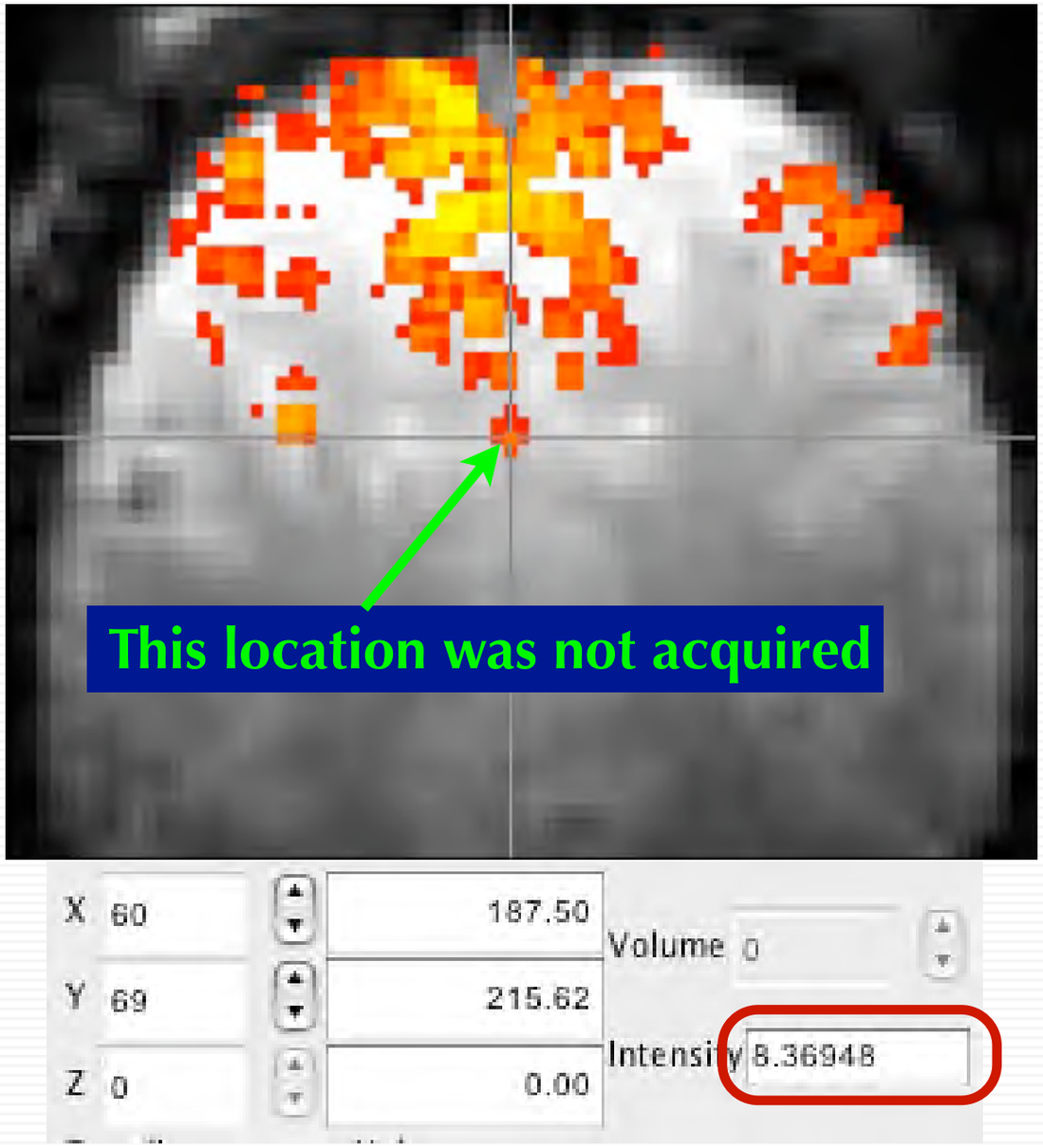
256 X 256

Imaging Time Held Constant

Interpolation



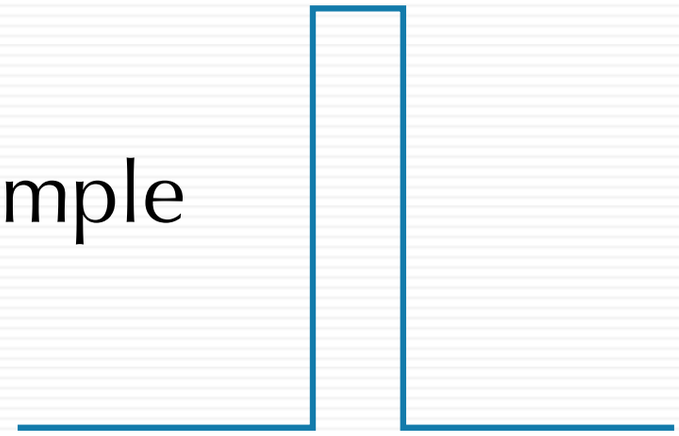
Native Resolution



Bilinear Interpolation

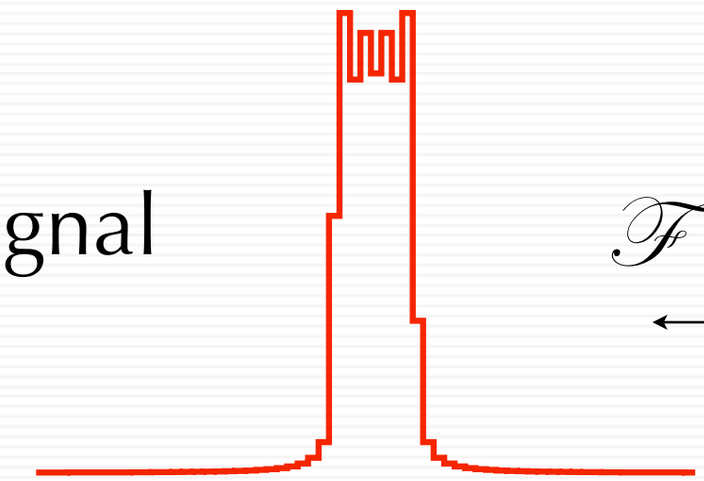
Truncation in Fourier Domain

Original Sample

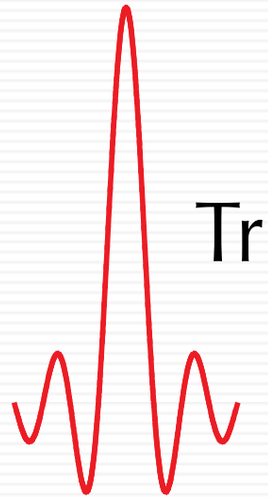


Frequency-encoded Signal

Apparent Signal

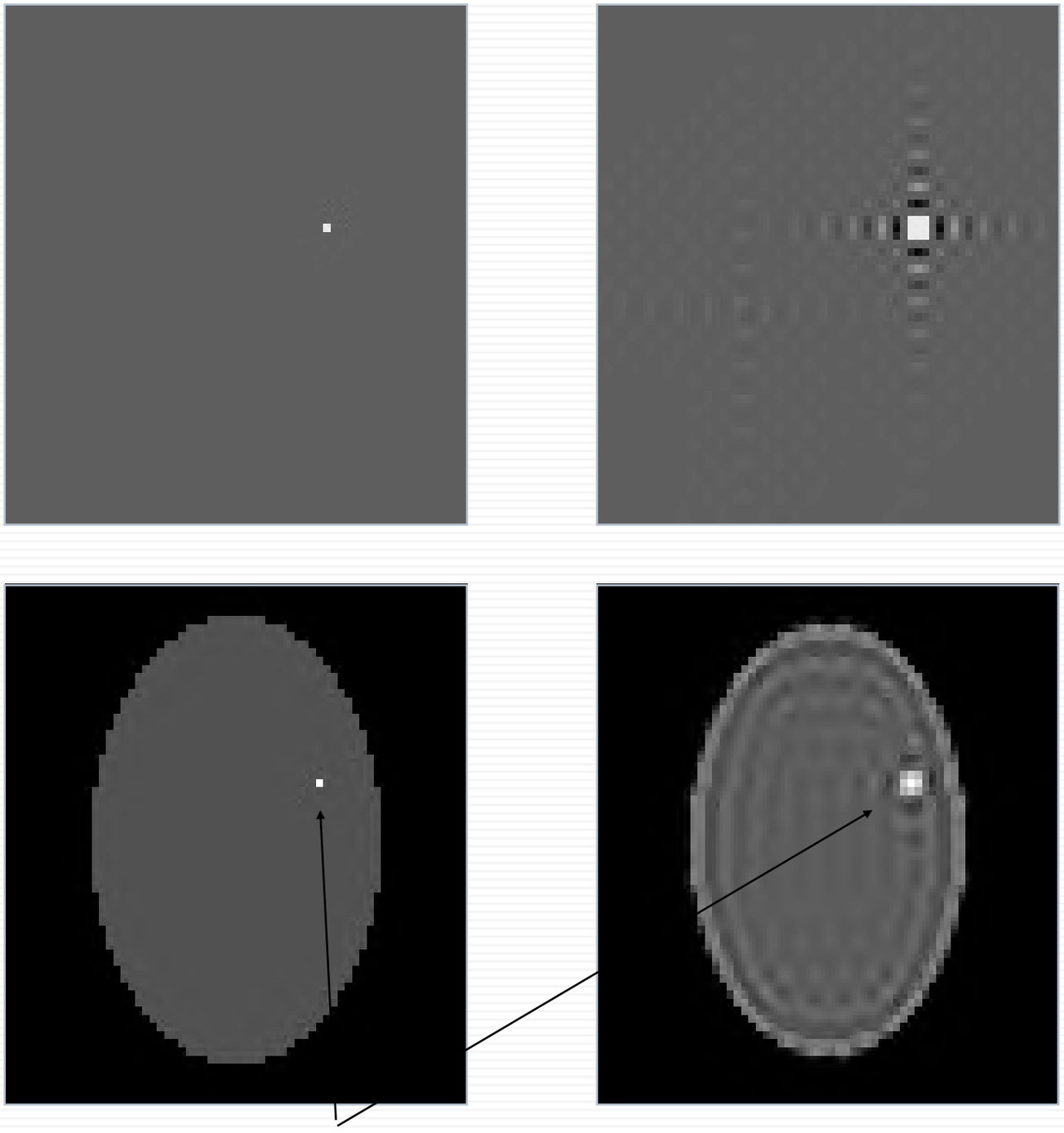


$\mathcal{F}^{-1}(s)$



Truncated Series

What is the actual resolution of MRI?

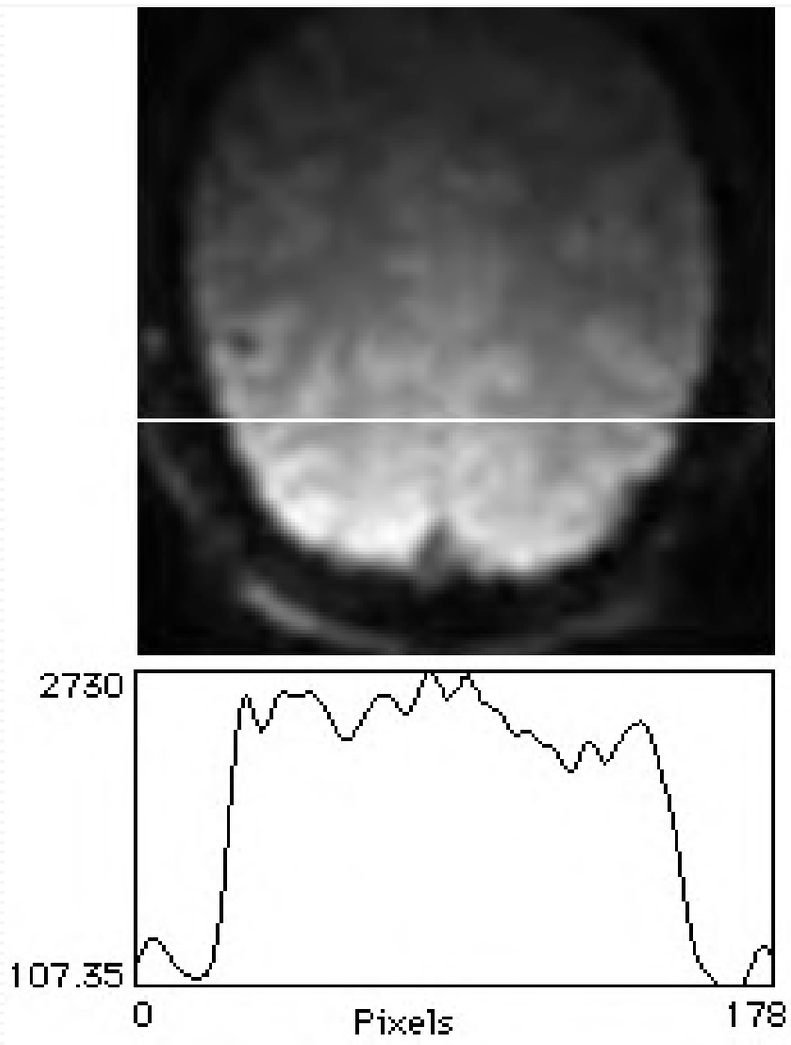


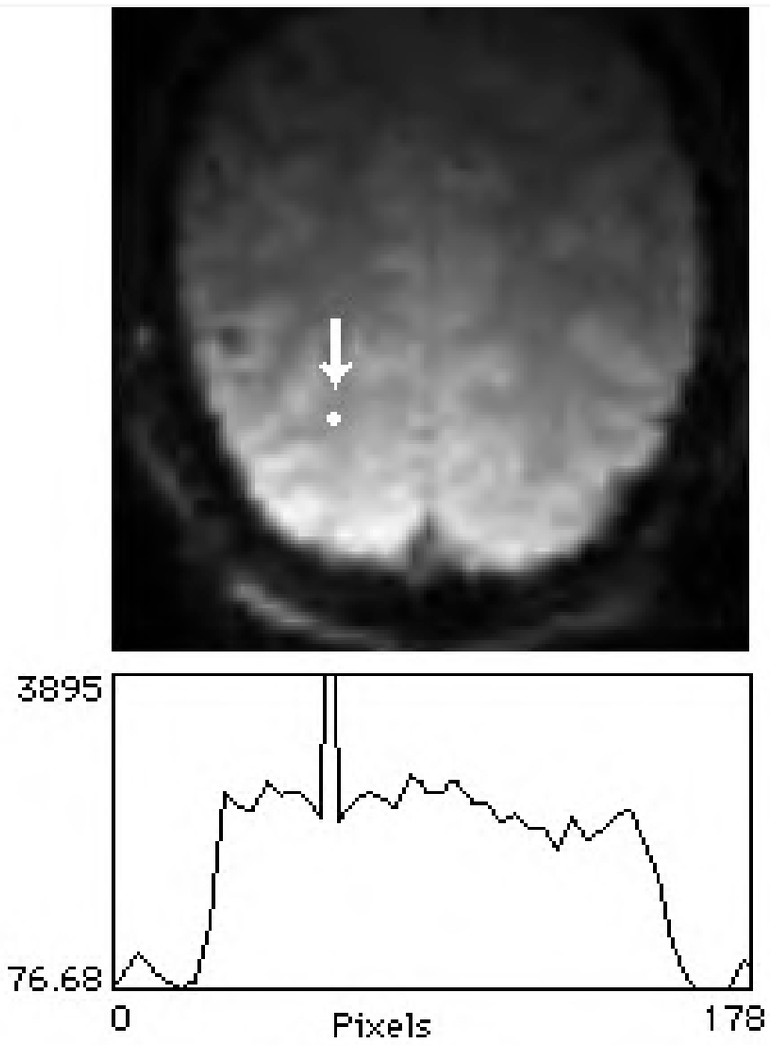
Original Data

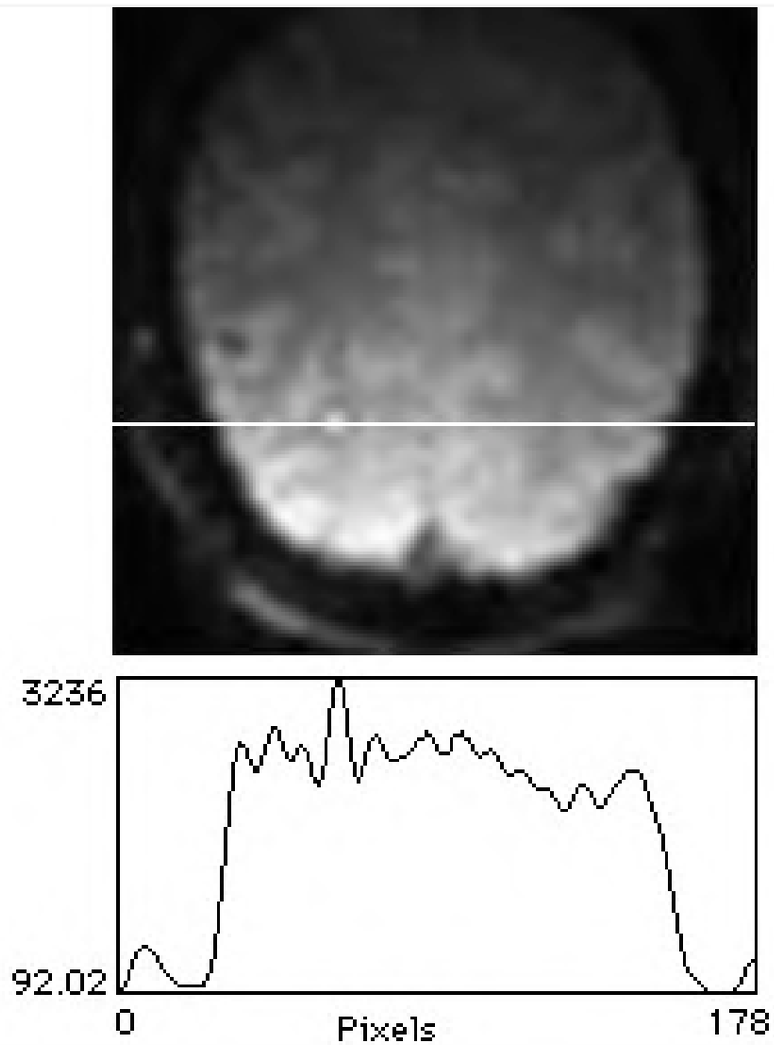
MR Image

Single pixel “activation”

The Actual Resolution of *f*MRI







http://ccn.ucla.edu/BMCweb/SharedCode/MRArtifacts/MRArtifacts.html

Bandwidth and Readout

- Position is encoded by FREQUENCY
- Bandwidth refers to the Frequency Difference from the center of the image to its edge:

$$\textit{Frequency per pixel} = \frac{2^{*}\,\textit{Bandwidth}}{\textit{number of pixels}} = \frac{1}{\textit{readout duration}}$$

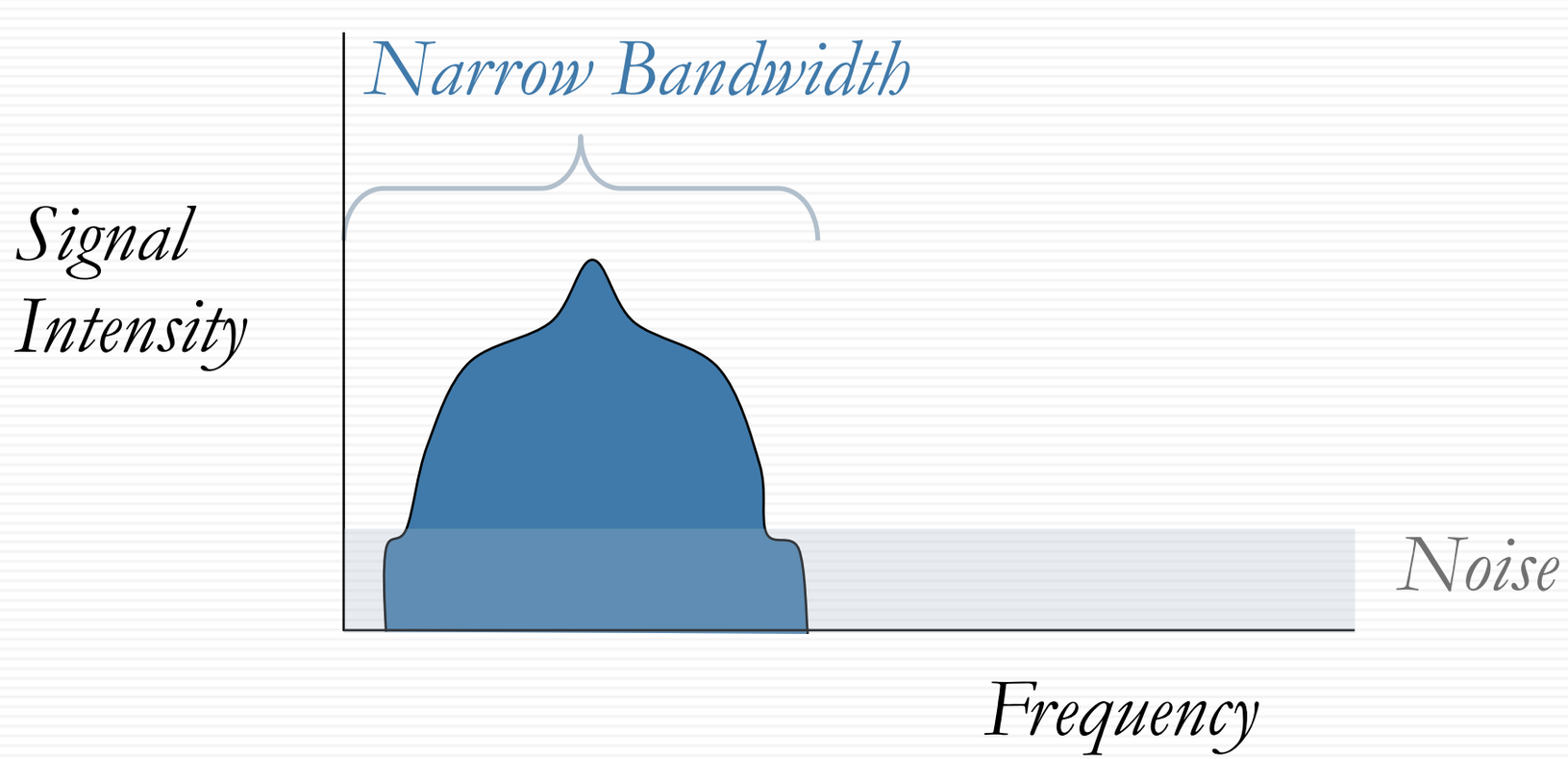
- Bandwidth decreases with readout duration:

$$\textit{Bandwidth} = \frac{\textit{number of pixels}}{2 * \textit{readout duration}}$$

Bandwidth and SNR

Decreasing the Bandwidth Improves SNR:

Imaging Time is INCREASED and high frequency noise is excluded



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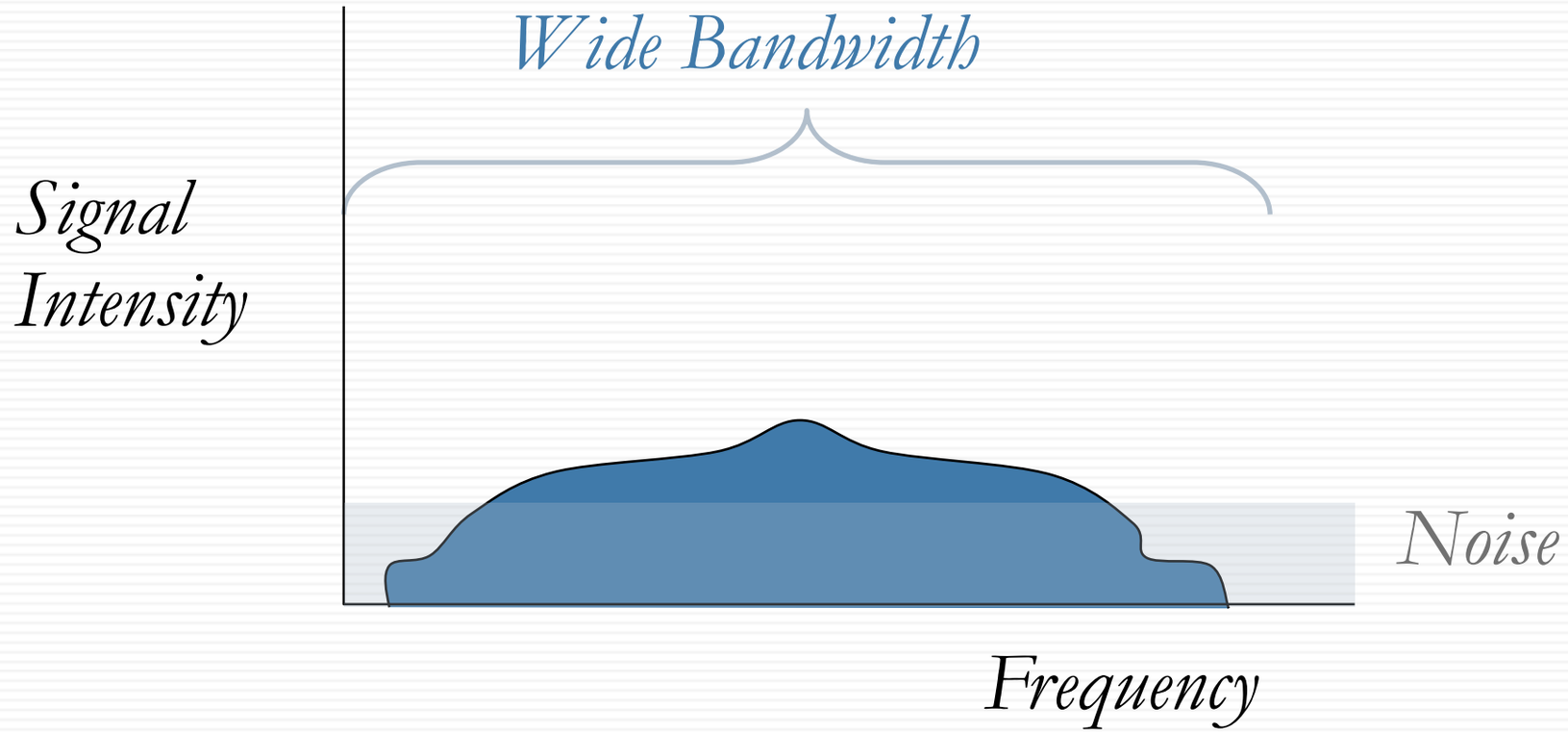
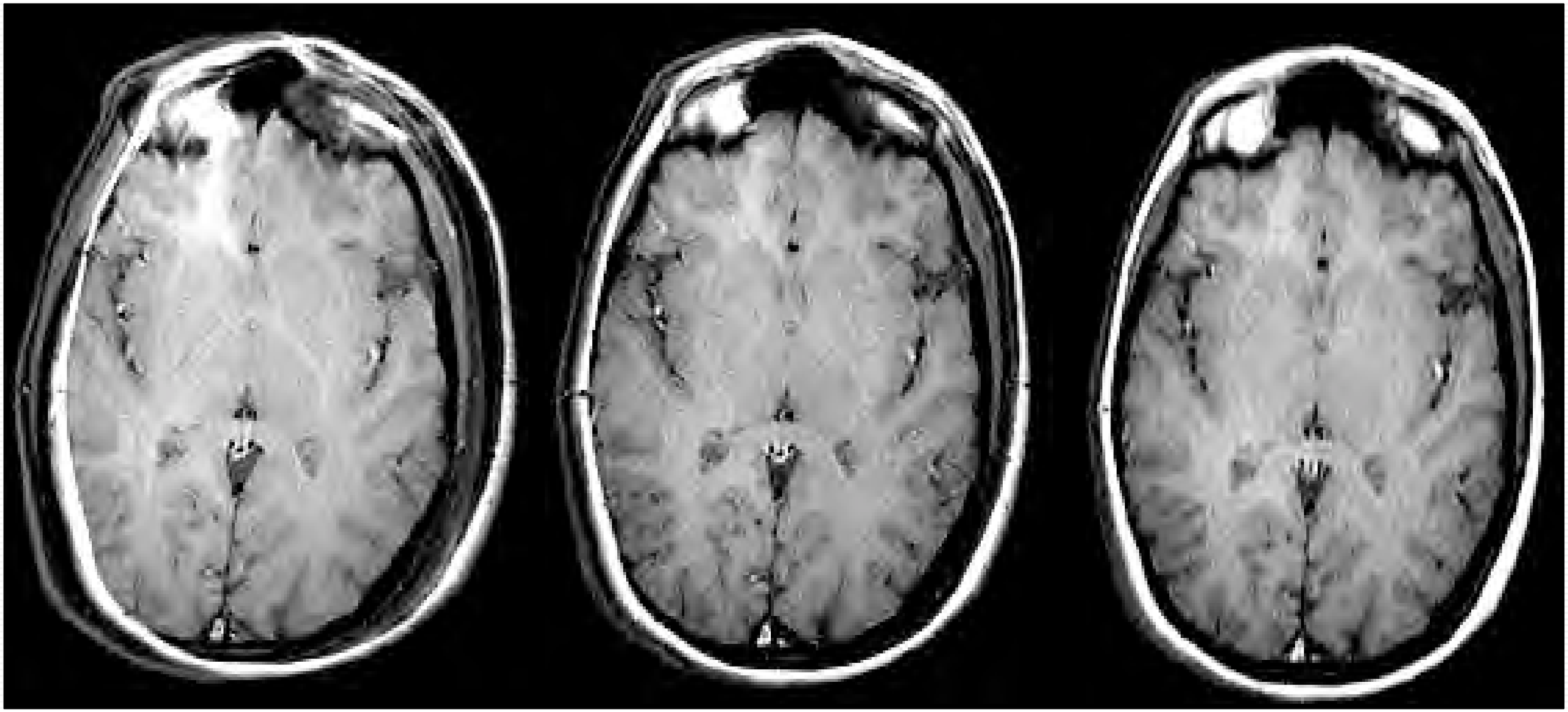


Bandwidth



BW=4kHz

BW=8kHz

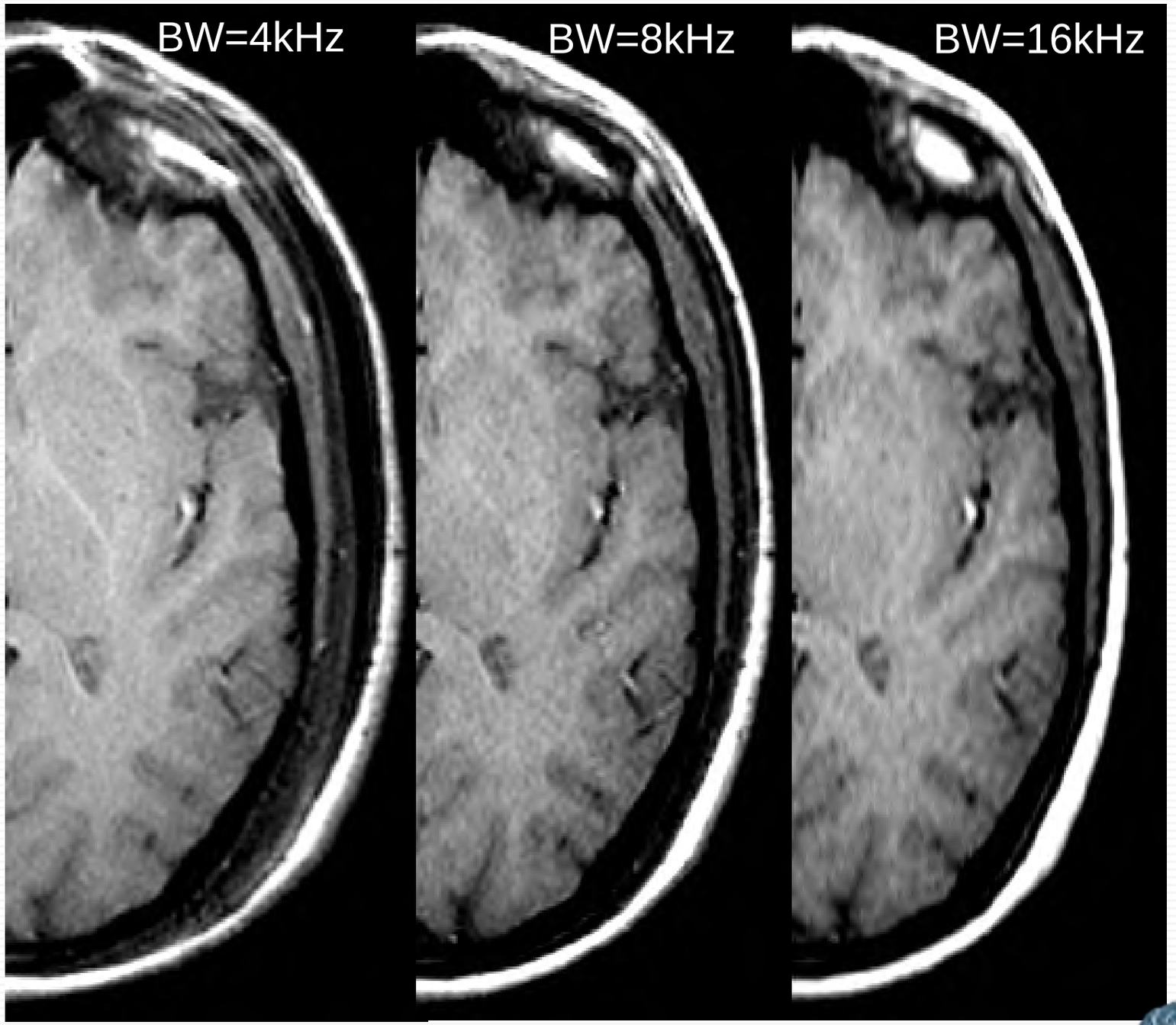
BW=16kHz

TE=11-14
TR=500

NEX=1
Matrix=256x256

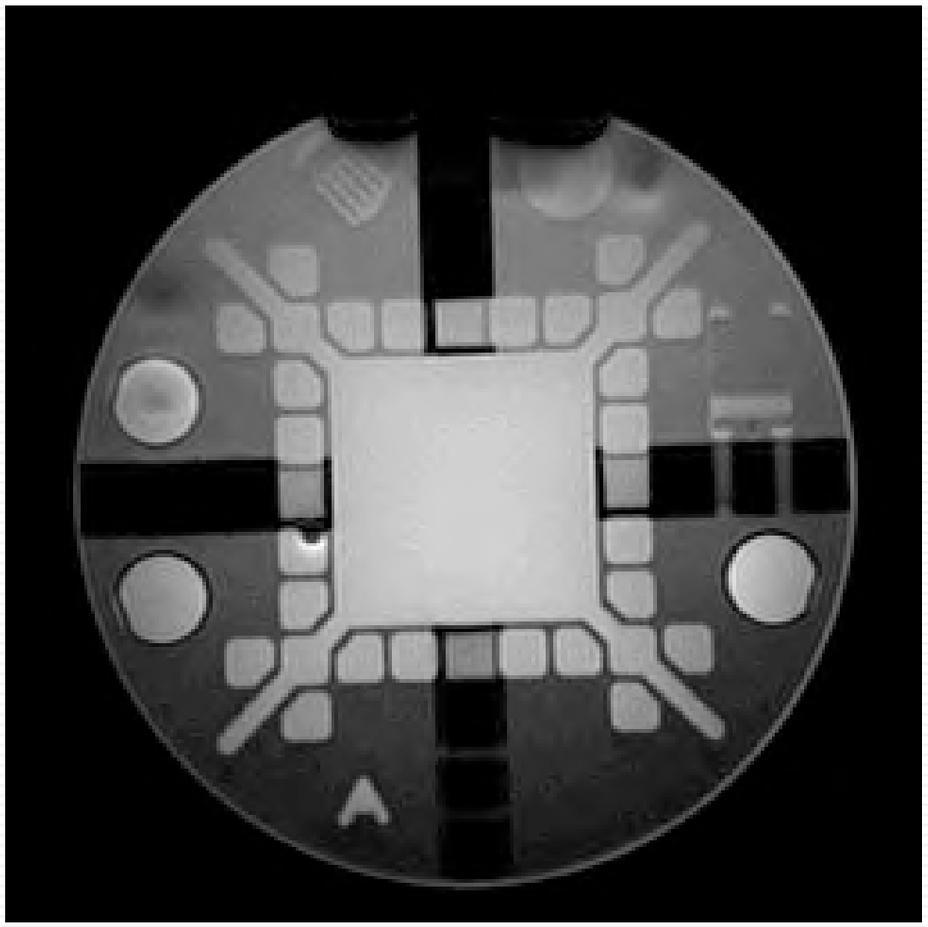
Thick=3mm

Bandwidth

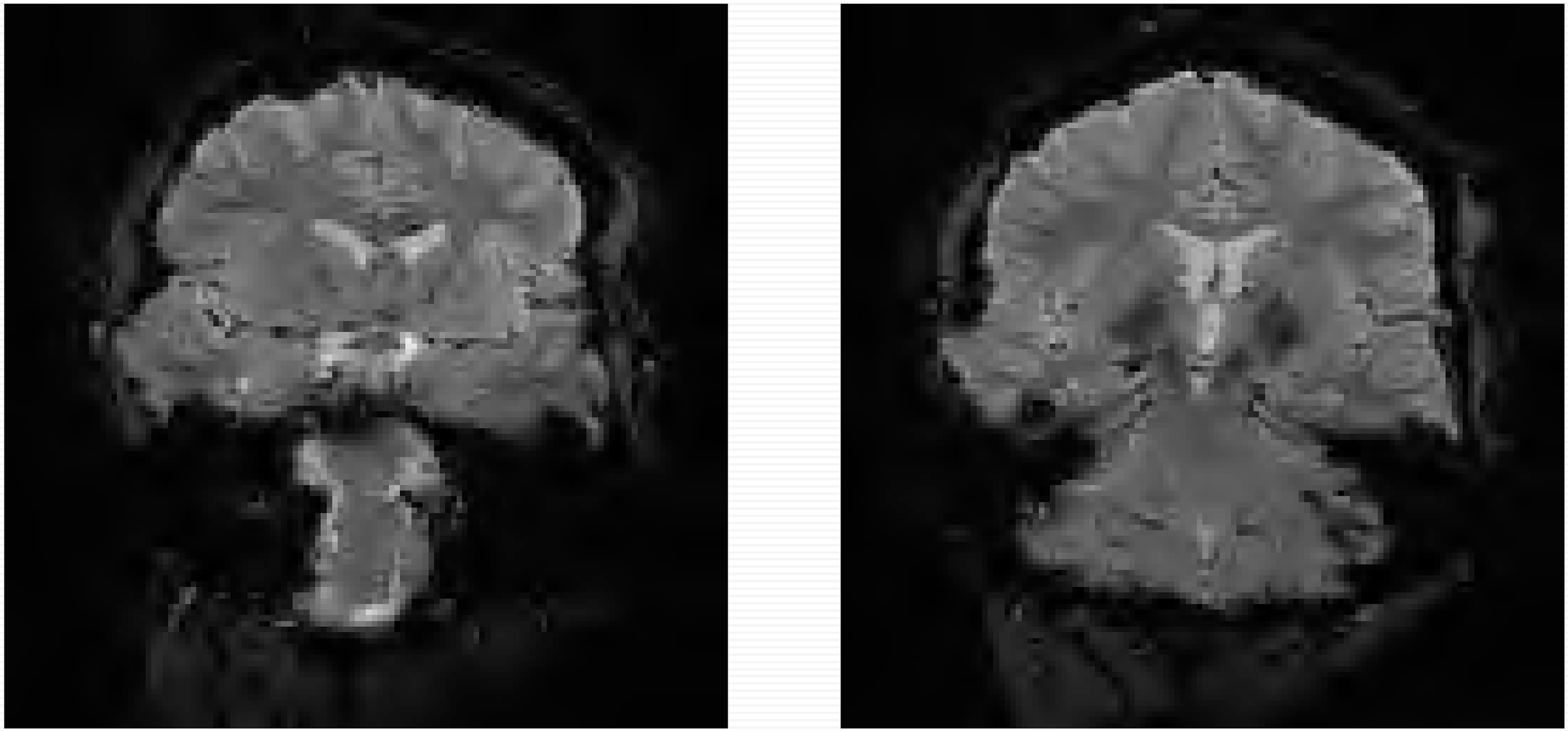


TE=11-14
NEX=1
Thick=3mm
TR=500
Matrix=256x256

Shape and Bandwidth



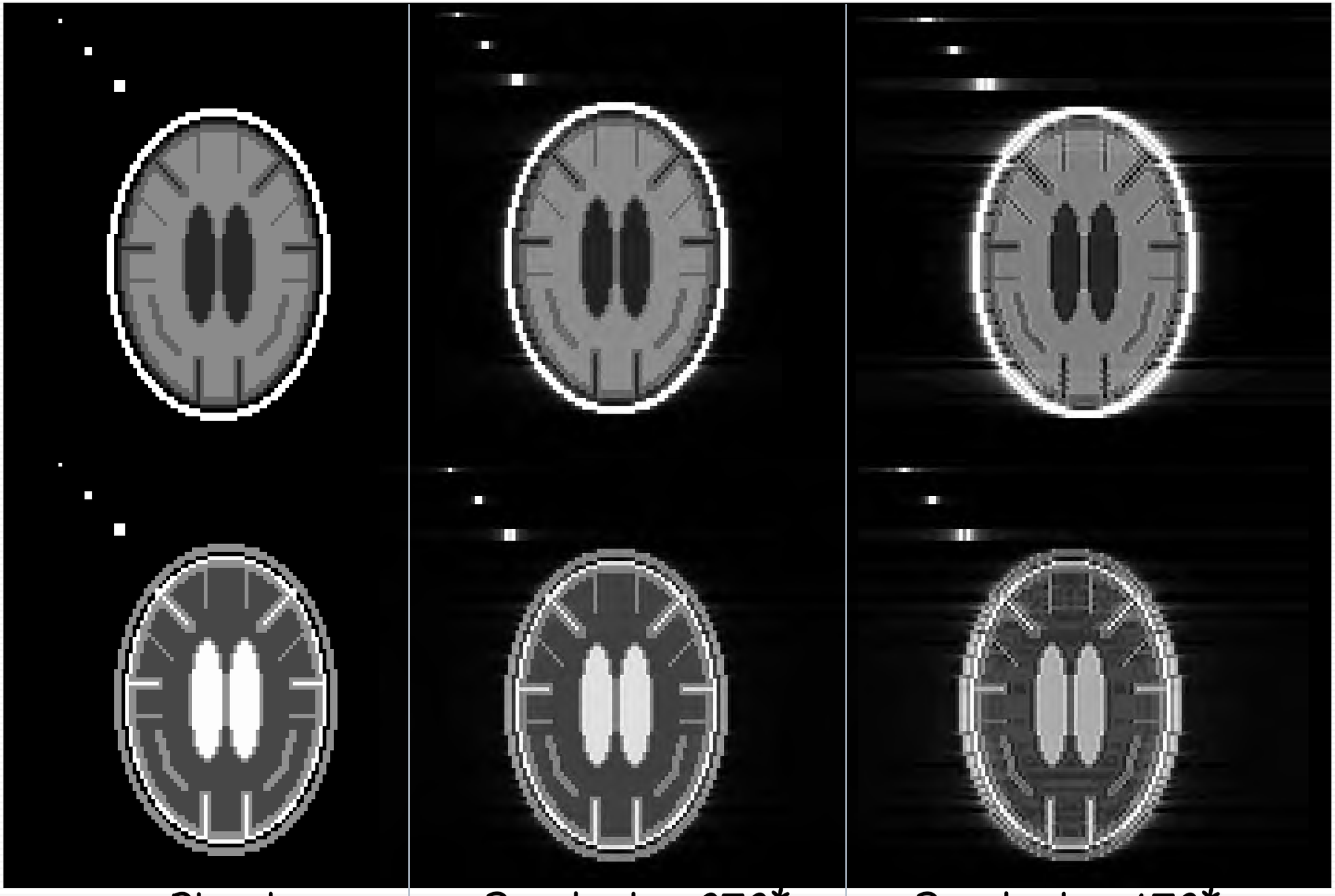
Distortions are More Severe at High Magnetic Field Strength



Variation in sample magnetization of is proportional to field strength.

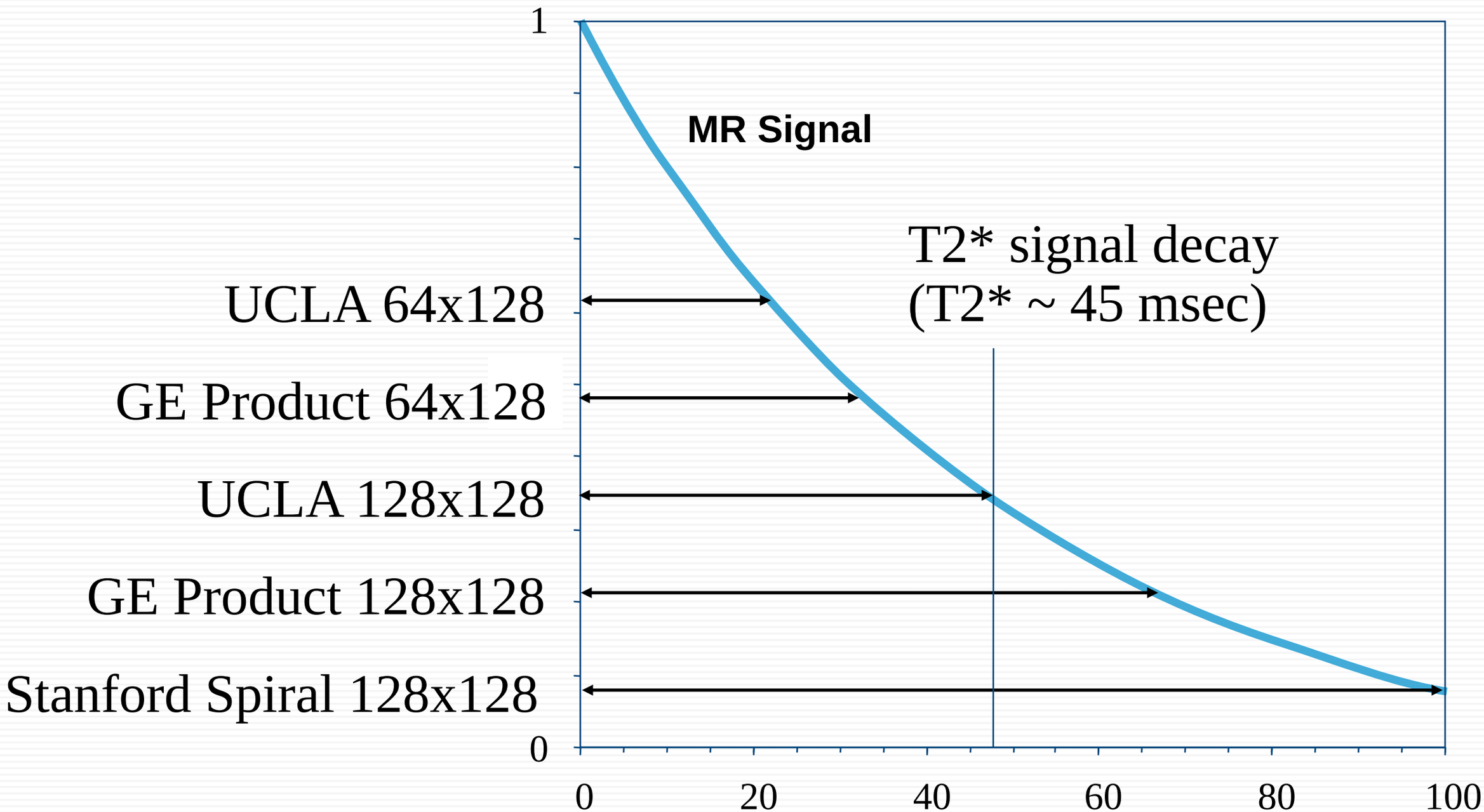
High Field images lose more signal from field inhomogeneity

Apodization from Long Readouts



Phantom | Readout = 2T2* | Readout = 4T2*

EPI Readout Durations

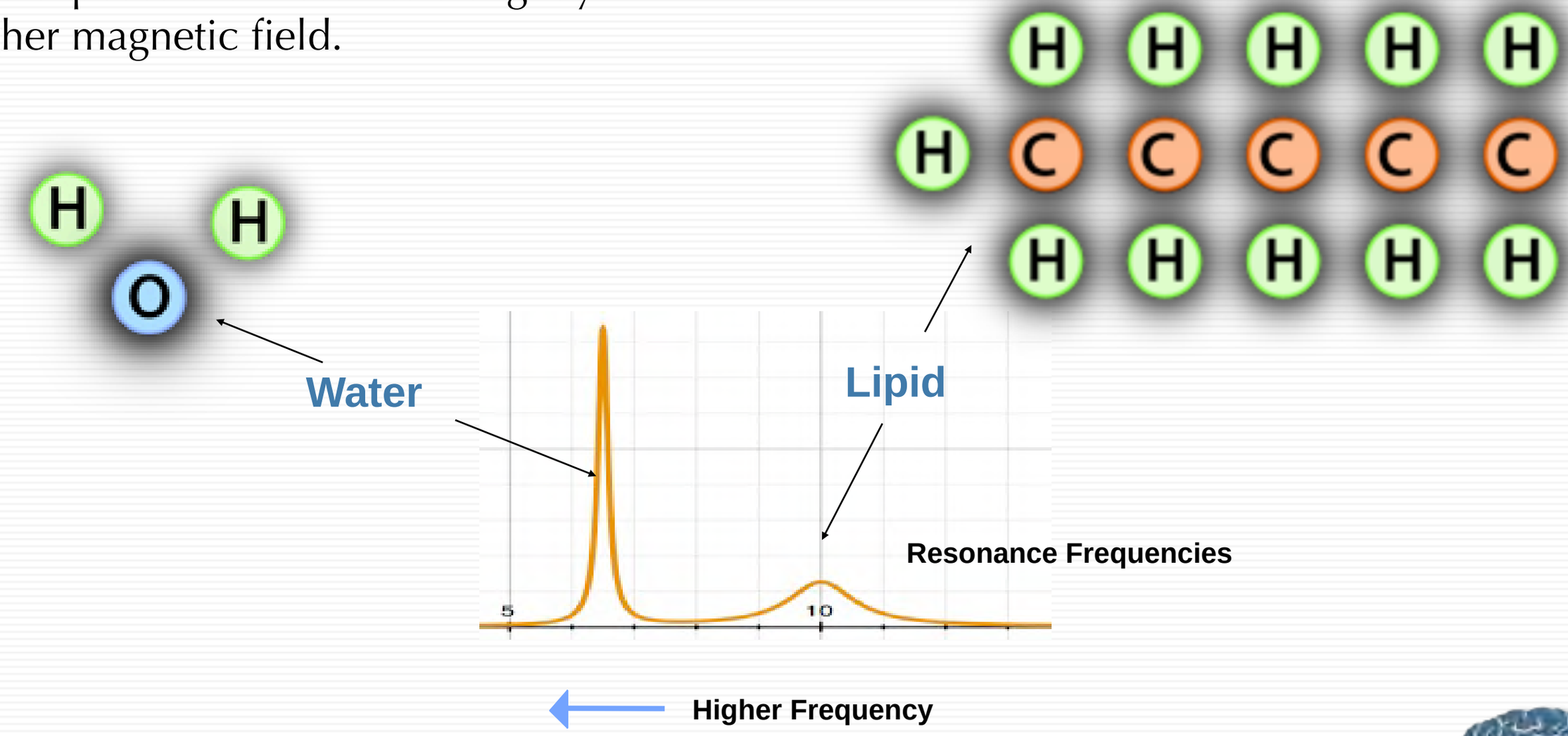


The Origin of Chemical Shift

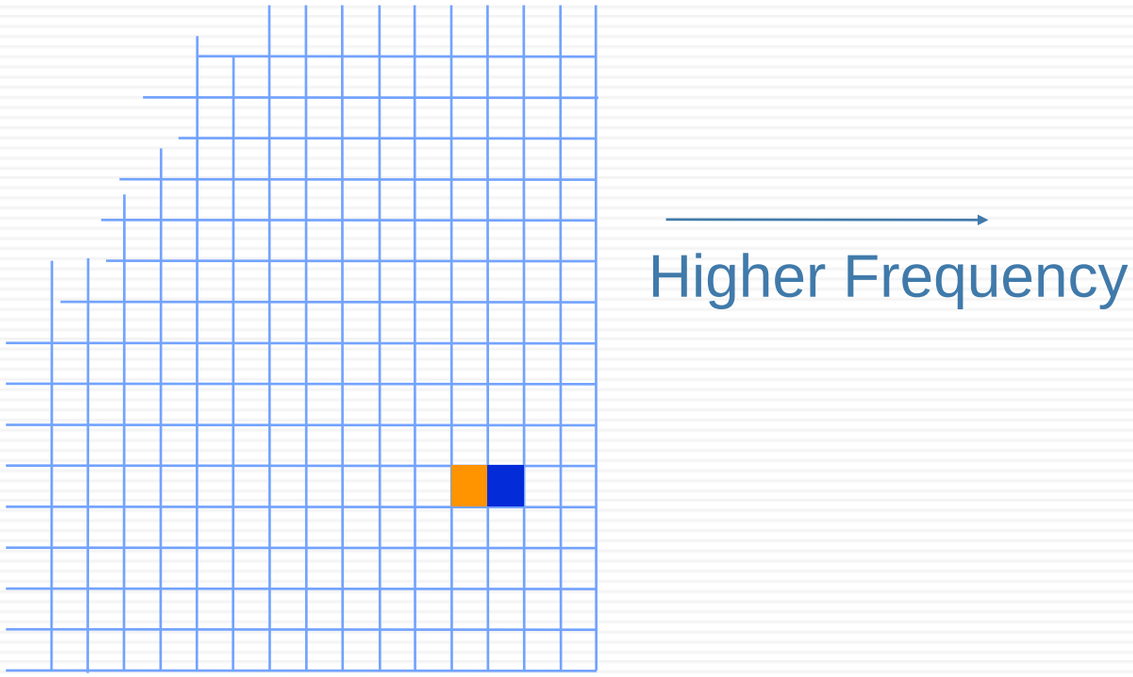
In water, electrons move from Hydrogen towards Oxygen.

This exposes the Proton to a slightly higher magnetic field.

Electrons in lipid are shared equally between Hydrogen and Oxygen



Chemical Shift Artifact

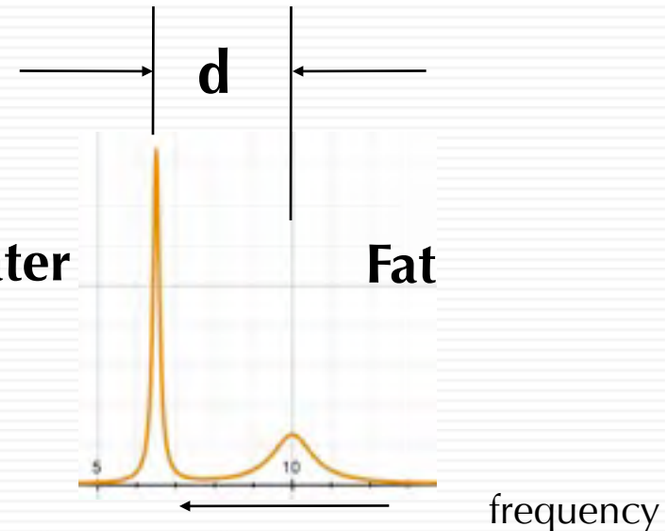


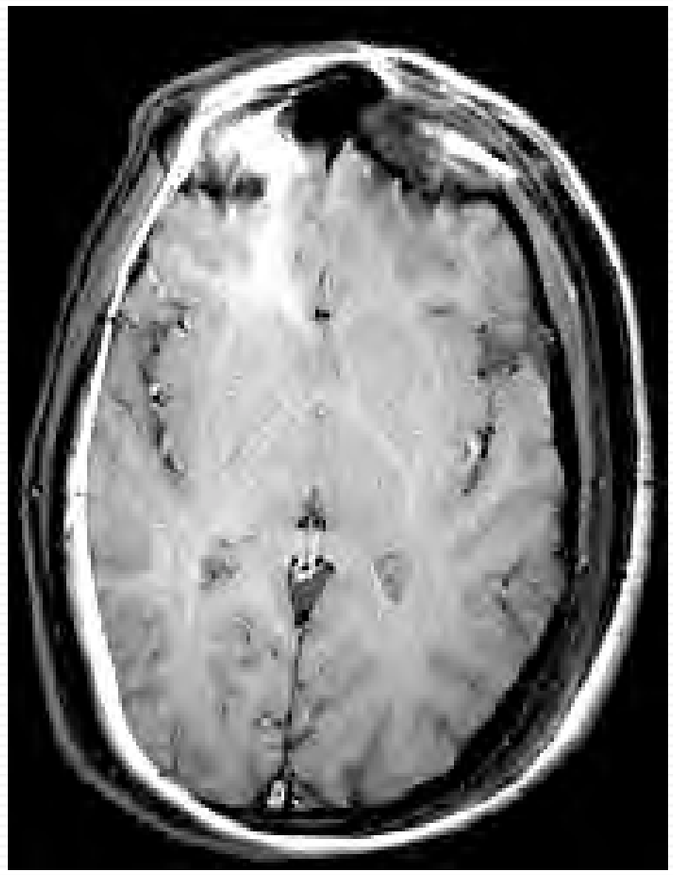
If the frequency *width* of each pixel is less than the frequency *difference* between water and lipid, then water and lipid will appear in separate pixels

Chemical Shift

The Fat-Water chemical shift is about 3.5 ppm or:

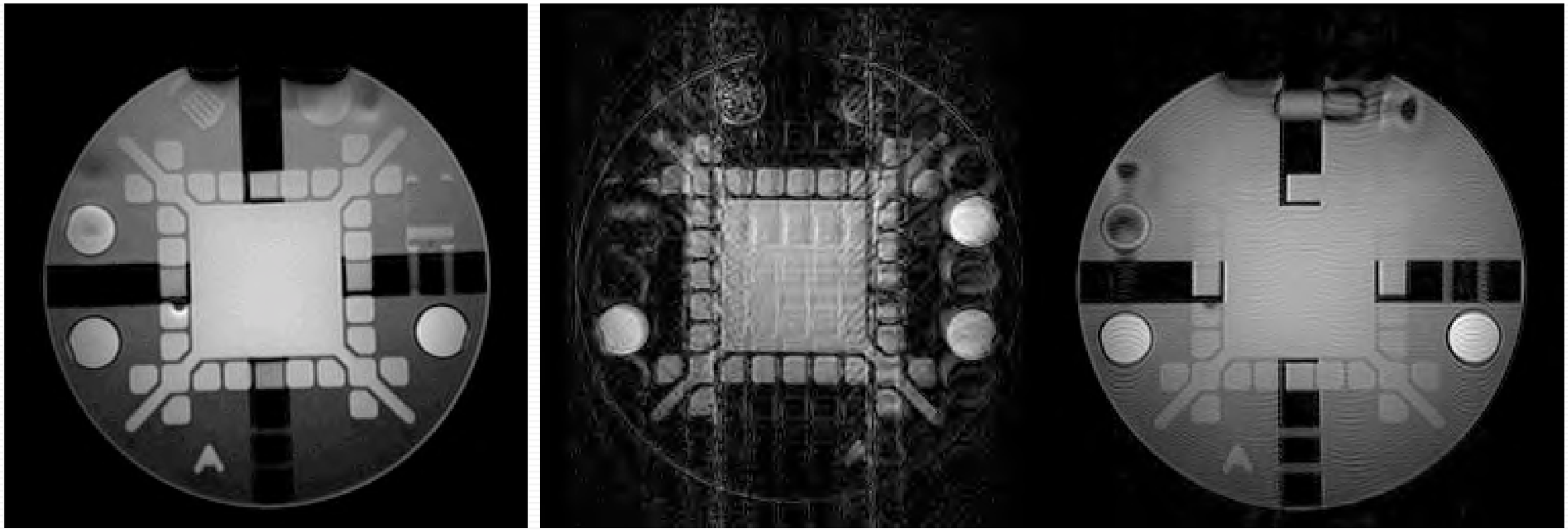
Which is:	*with a 32 kHz readout*
75 Hz @ 0.5 Tesla	< 1 pixel
150 Hz @ 1.0 Tesla	≈ 1 pixel
220 Hz @ 1.5 Tesla	> 1 pixel
440 Hz @ 3.0 Tesla	≈ 3.5 pixels





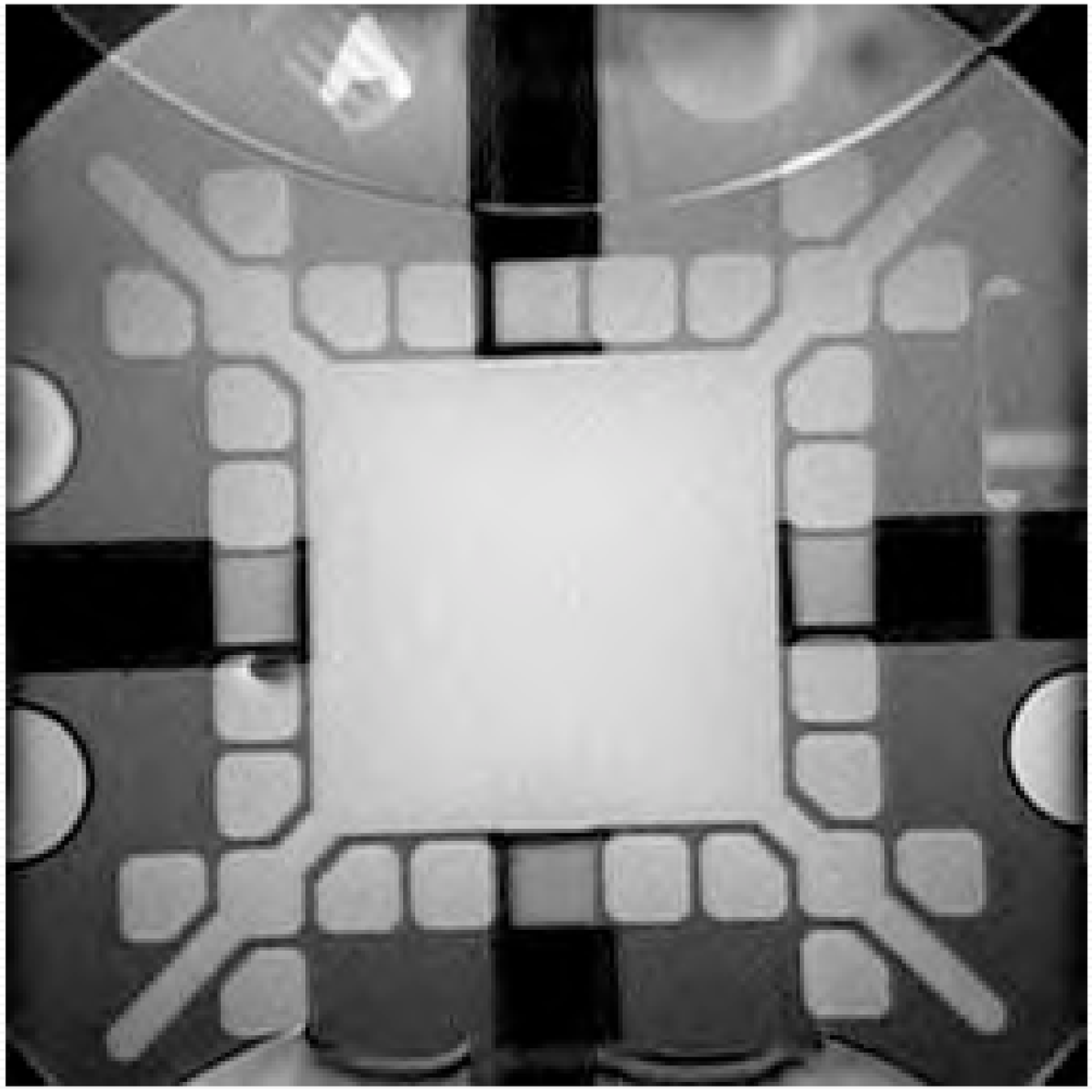
Lowering the Bandwidth/pixel increases the Chemical Shift in pixels

Motion Artifact

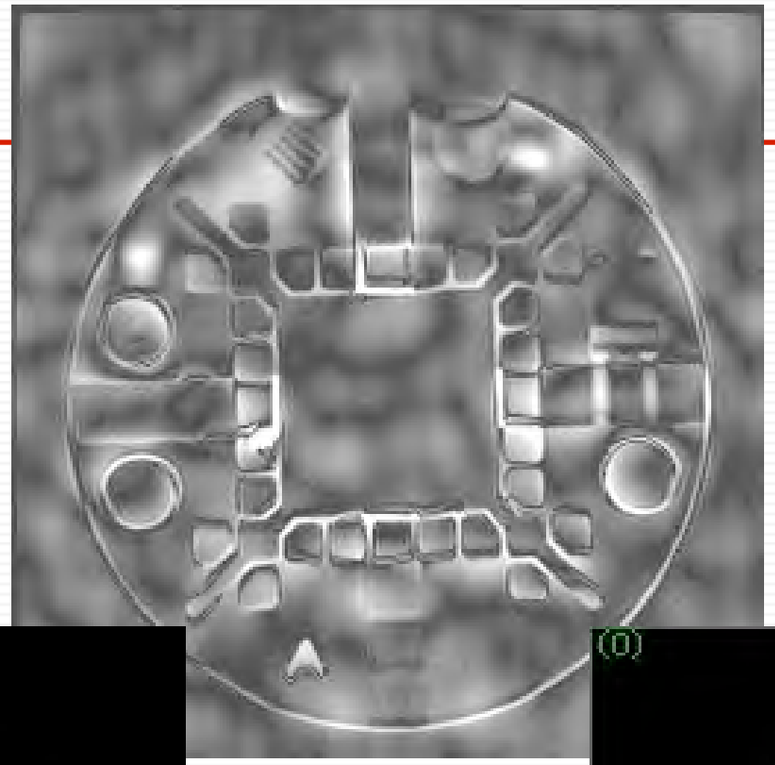


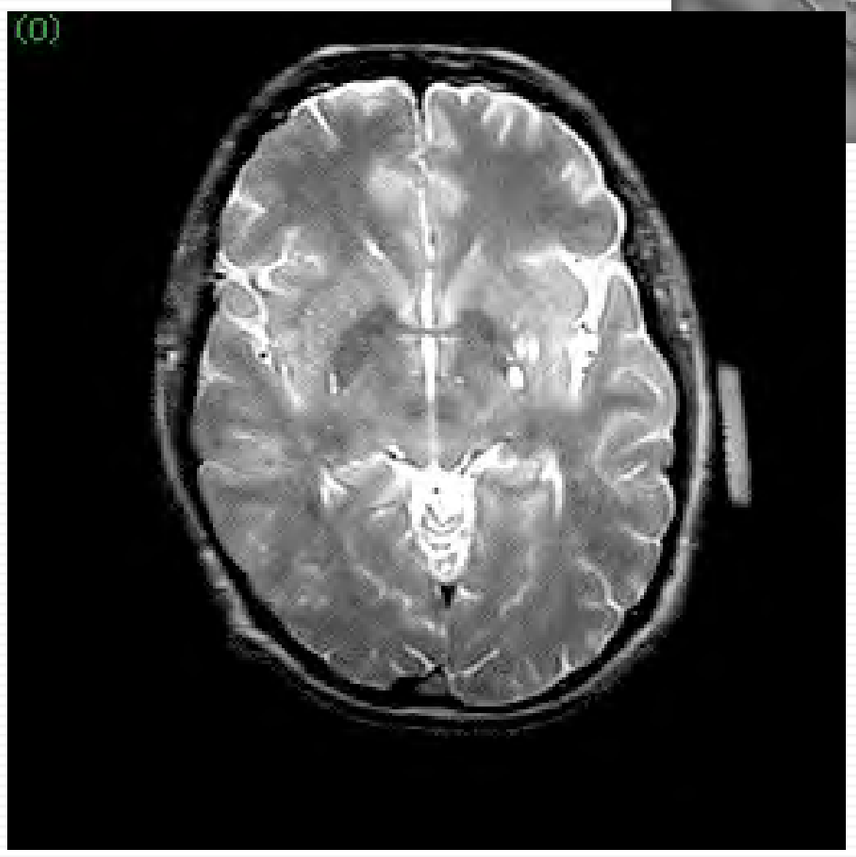
http://airto.ccn.ucla.edu/BMCweb/SharedCode/MRArtifacts/MRArtifacts.html

Aliasing

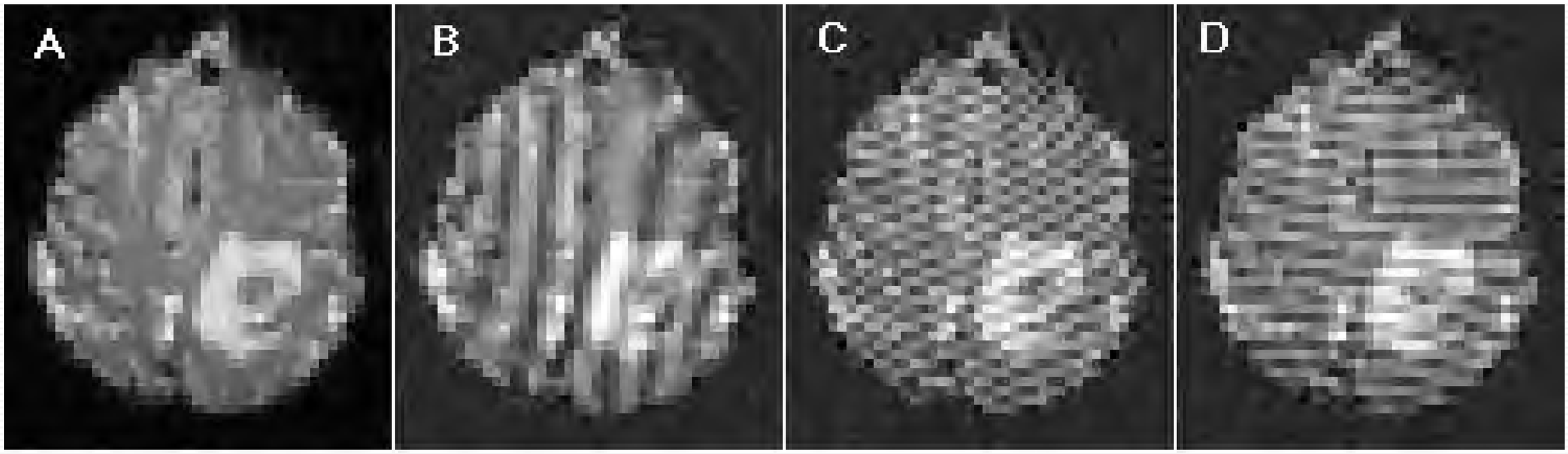


Saturation

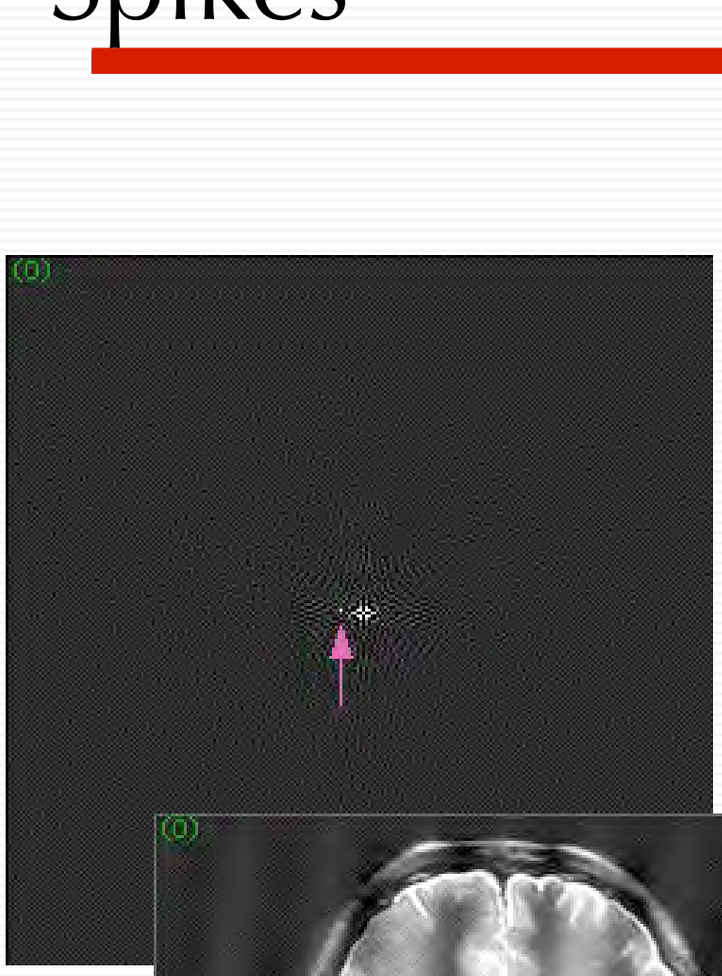


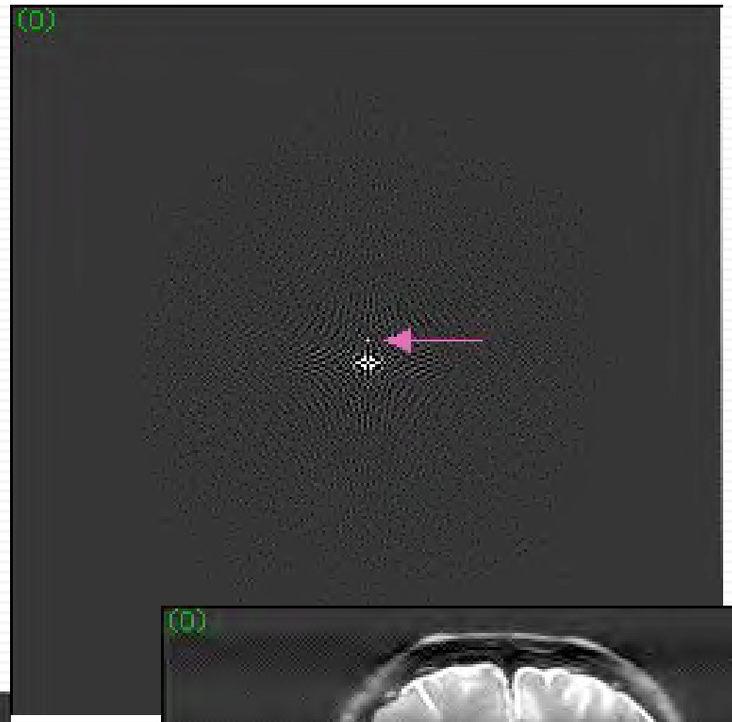


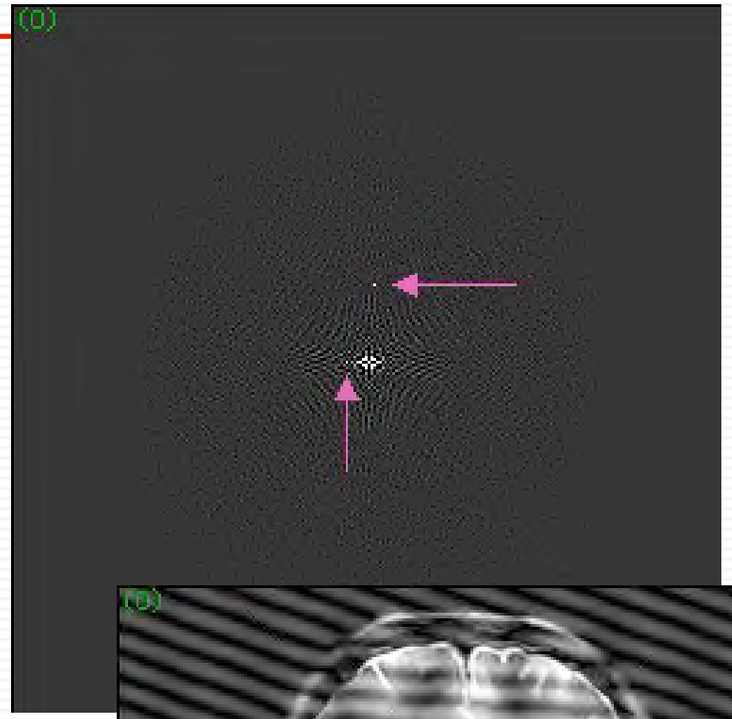
Spikes

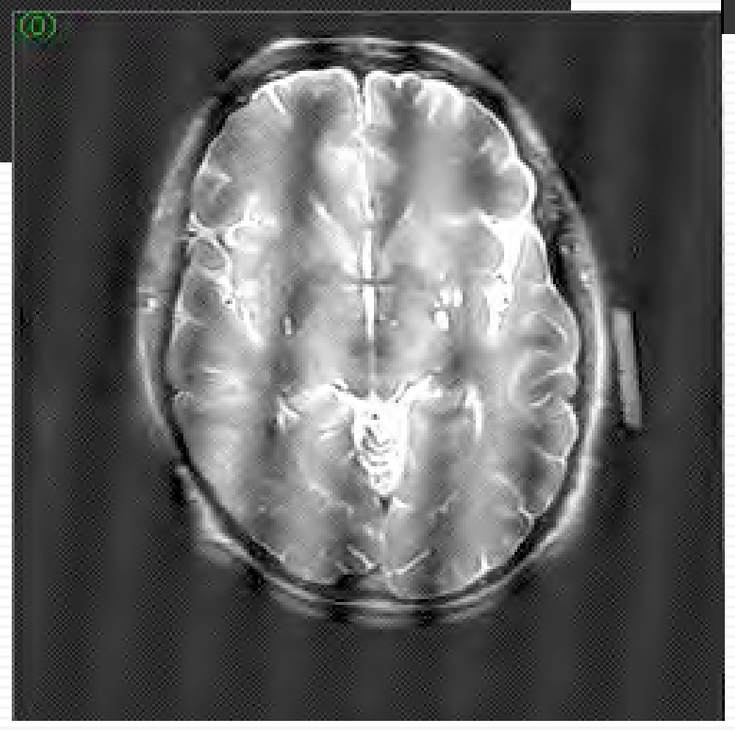


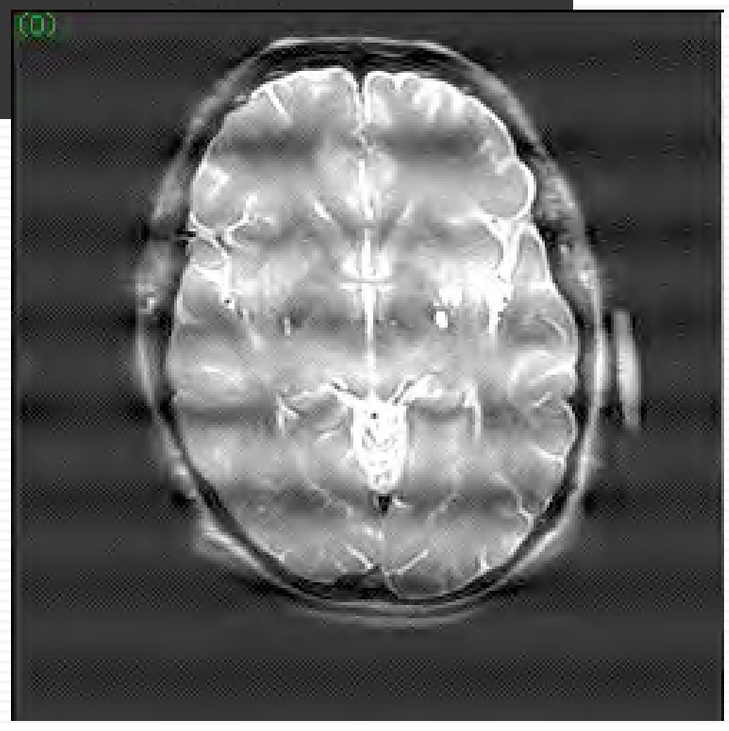
Spikes











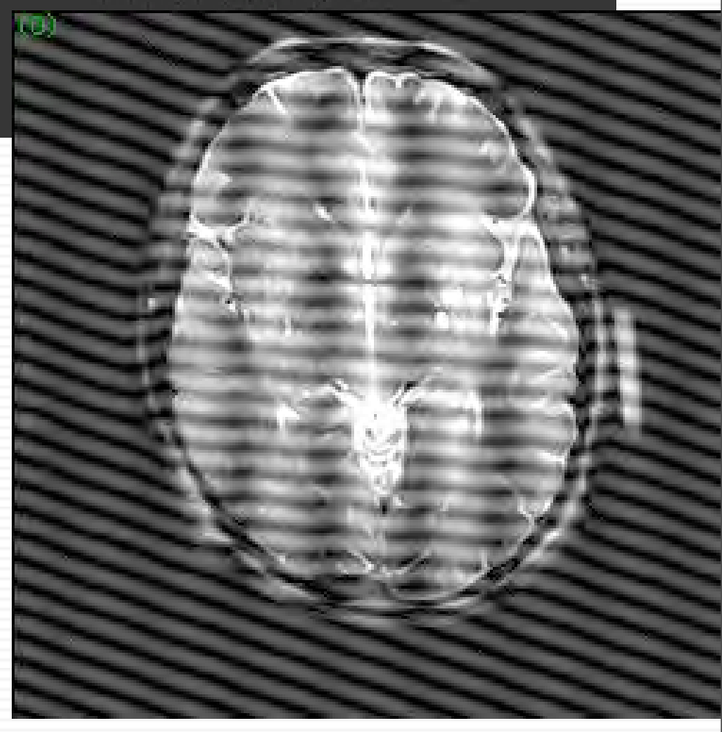


Image Quality

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- SNR is Very Limited in MRI

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- Feature Detection Falls Rapidly with Loss in Contrast to Noise Ratio

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Image Quality

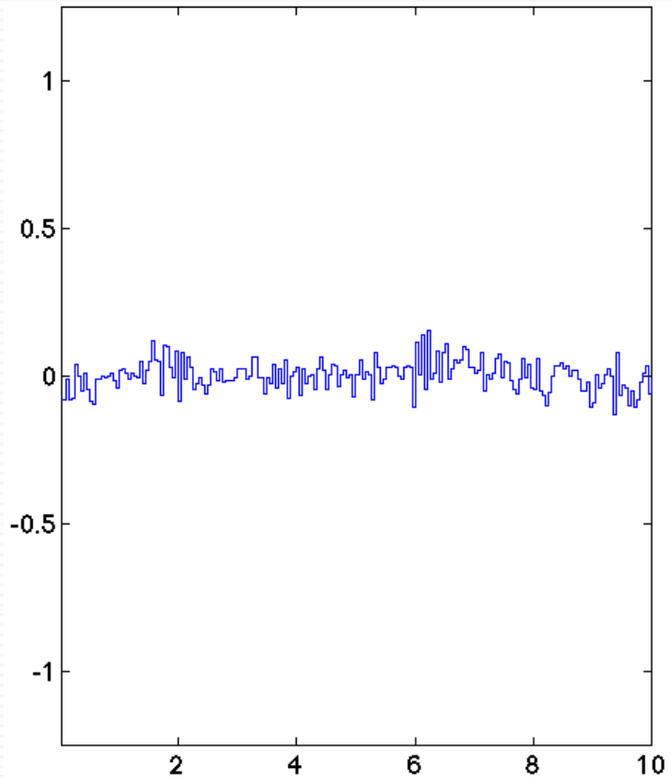
- SNR is Very Limited in MRI
- Feature Detection Falls Rapidly with Loss in Contrast to Noise Ratio
- Usable Resolution is NOT the Same as Voxel Size
- Spatial Encoding Artifacts in MRI May Have Complex Appearance
- Edge Ringing and Blurring are Related to Parameters Such as Contrast

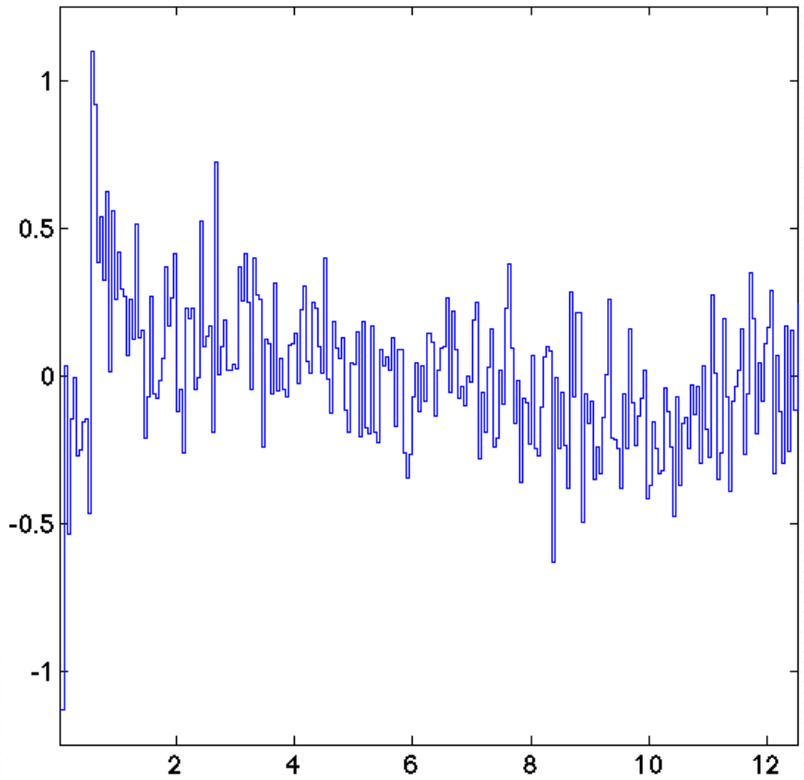
Characterize Your Tools

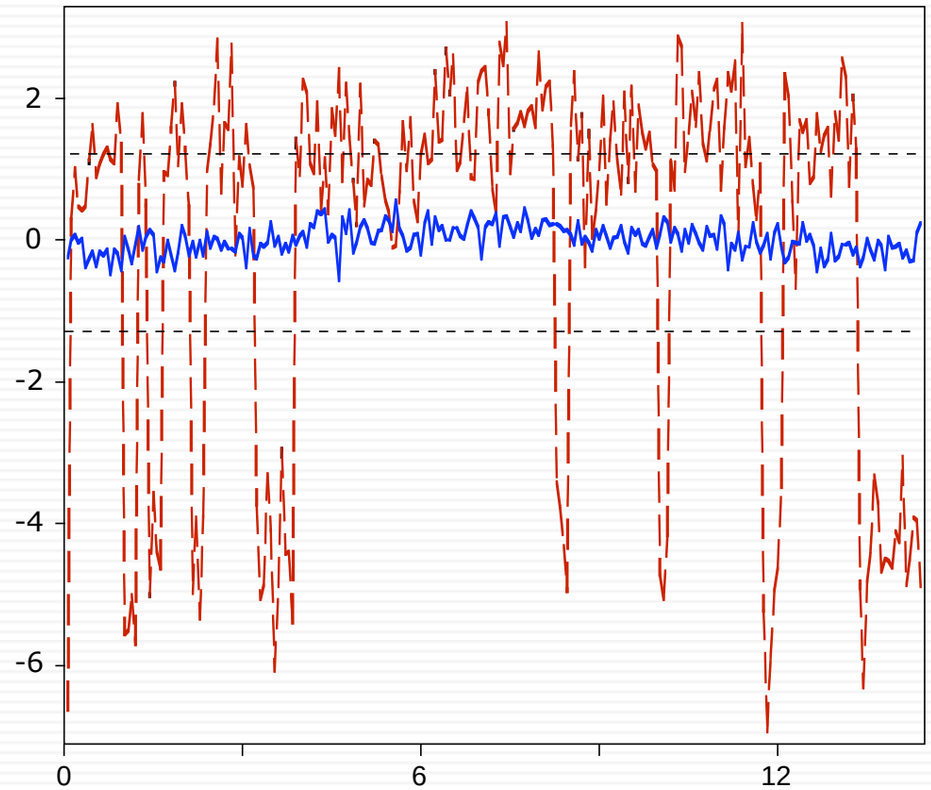
- Test Statistics are Effect/Variance
- Variance includes:
  - Intrasubject (motion, attention, physiology, fatigue,…)
  - Intersubject variance (position, morphology, performance, pathology, physiology,…)
  - Experimental Variance (uncontrolled variables, stimulation variance,…)
  - Instrument Variance
  - Sitewise Variance
  - True Random Noise

Instrument Variation

1. System Instability

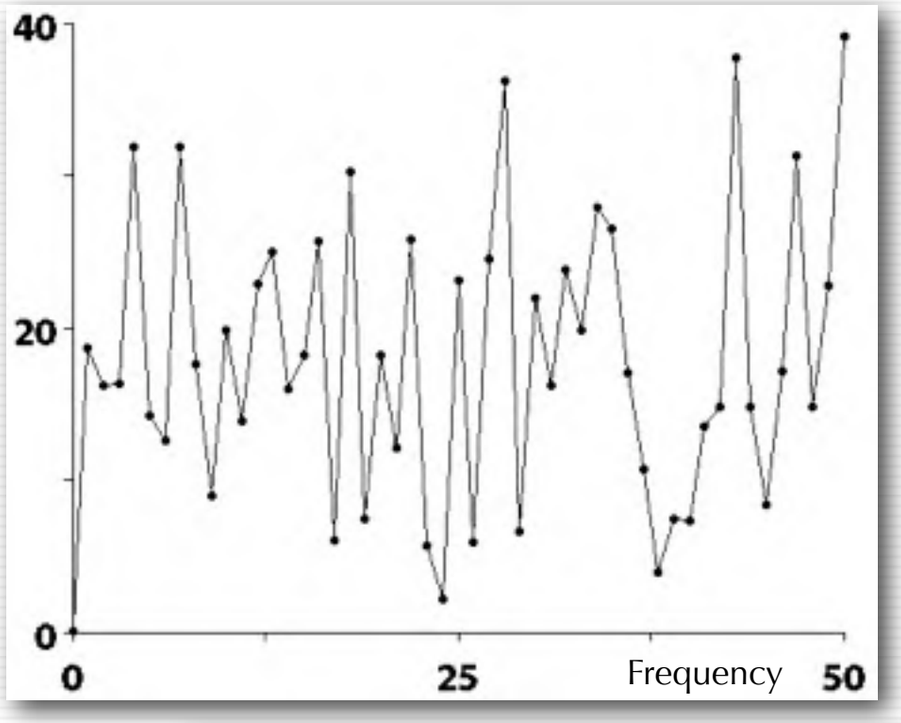


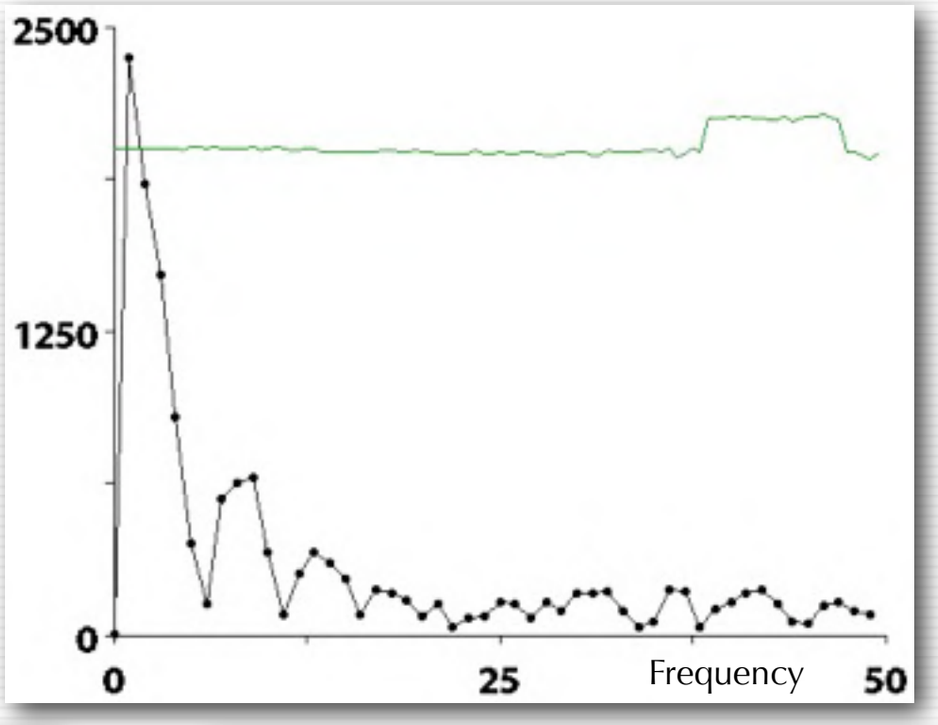


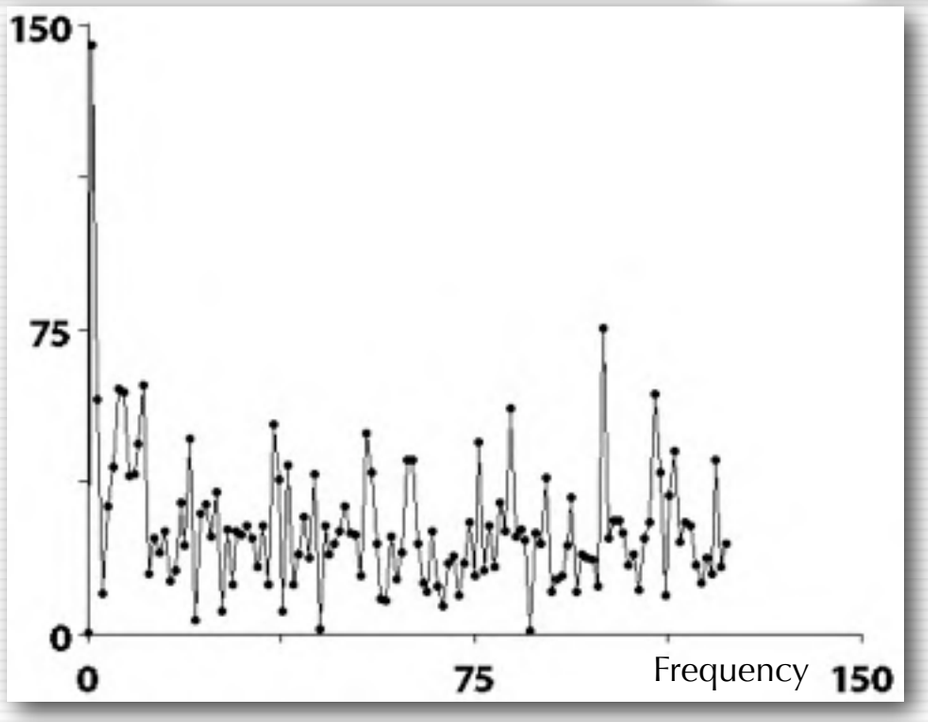


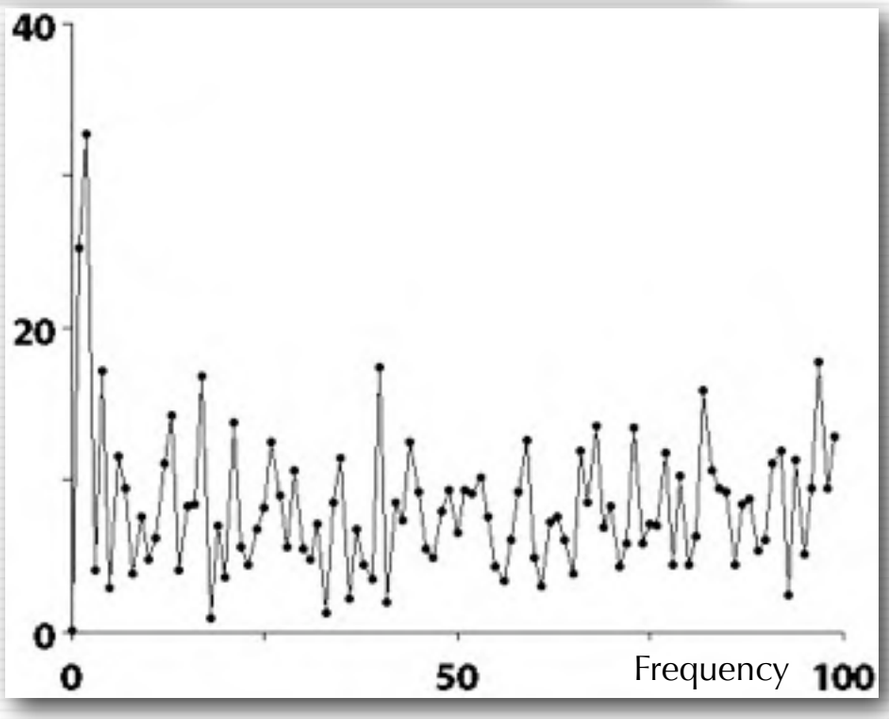
Mean Intensity Variation

Scanner Autocorrelation

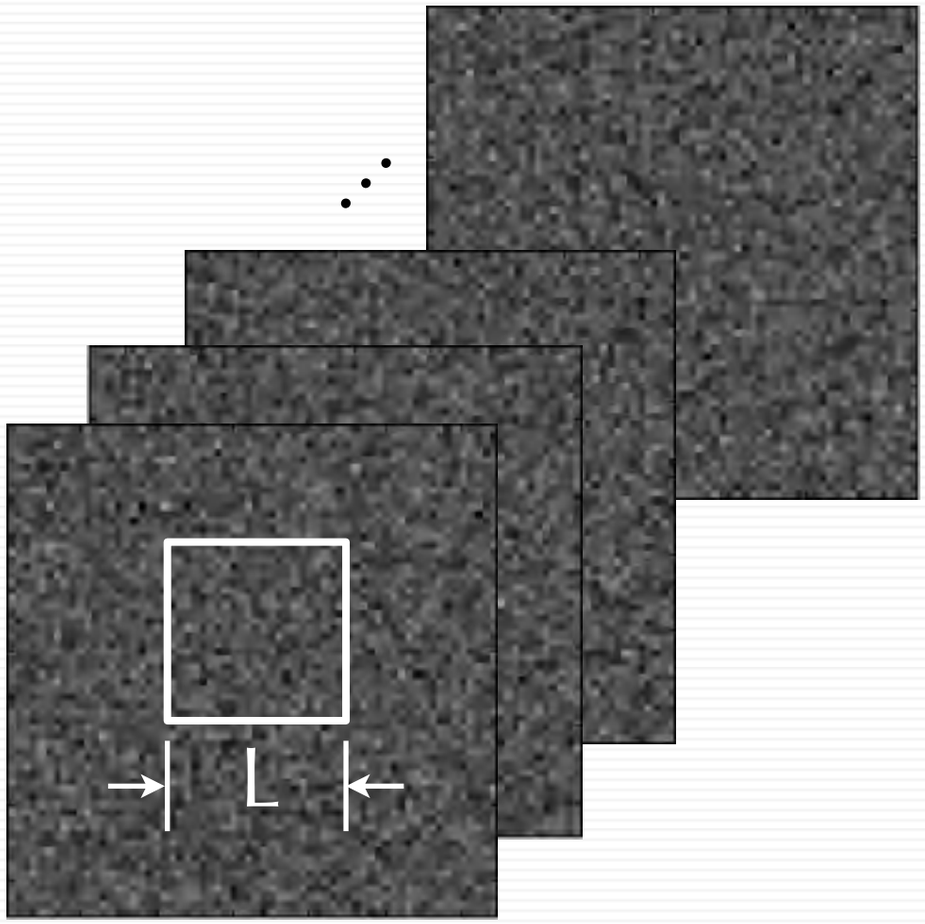




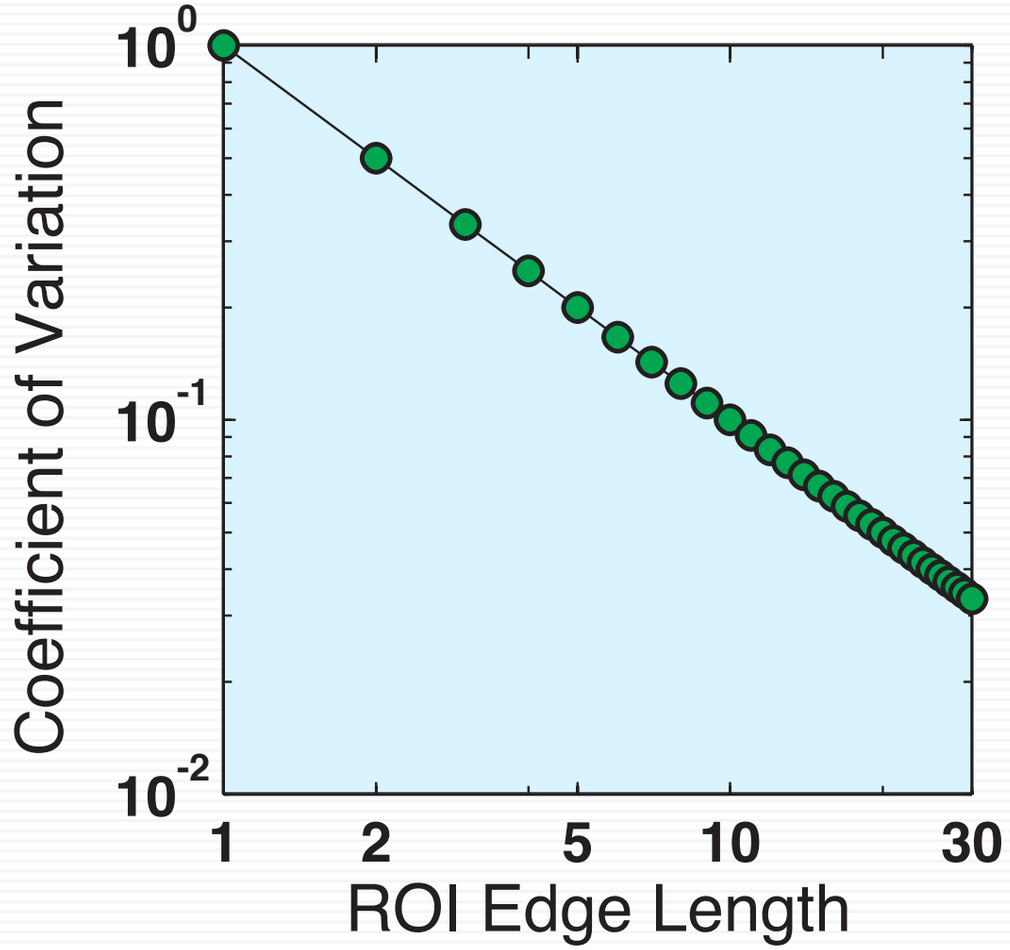




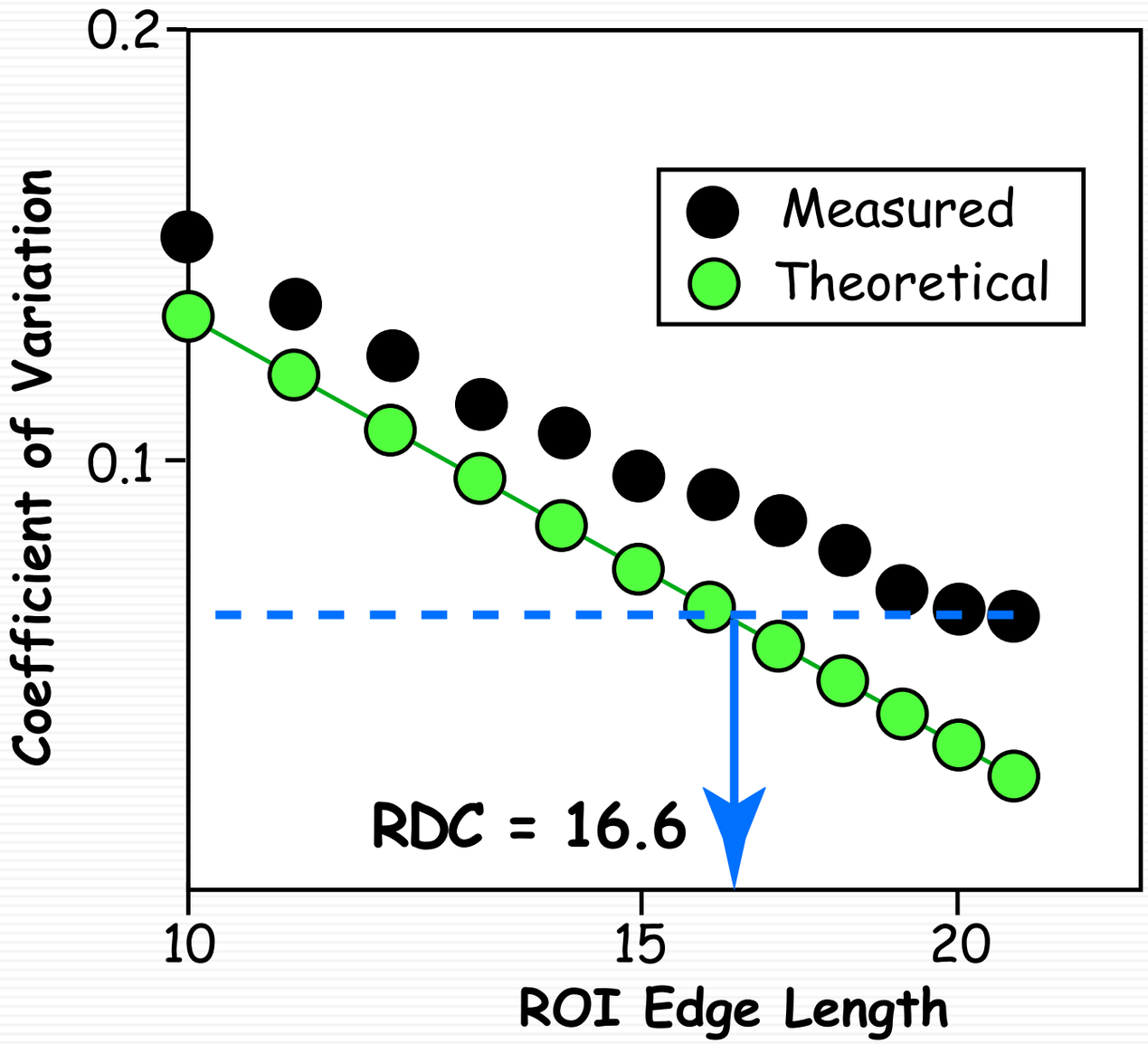
the Weisskoff Plot



The Expected Standard Deviation of the Mean Signal of a Region over Time Falls with the Square Root of the Number of Voxels.



the Weisskoff Plot

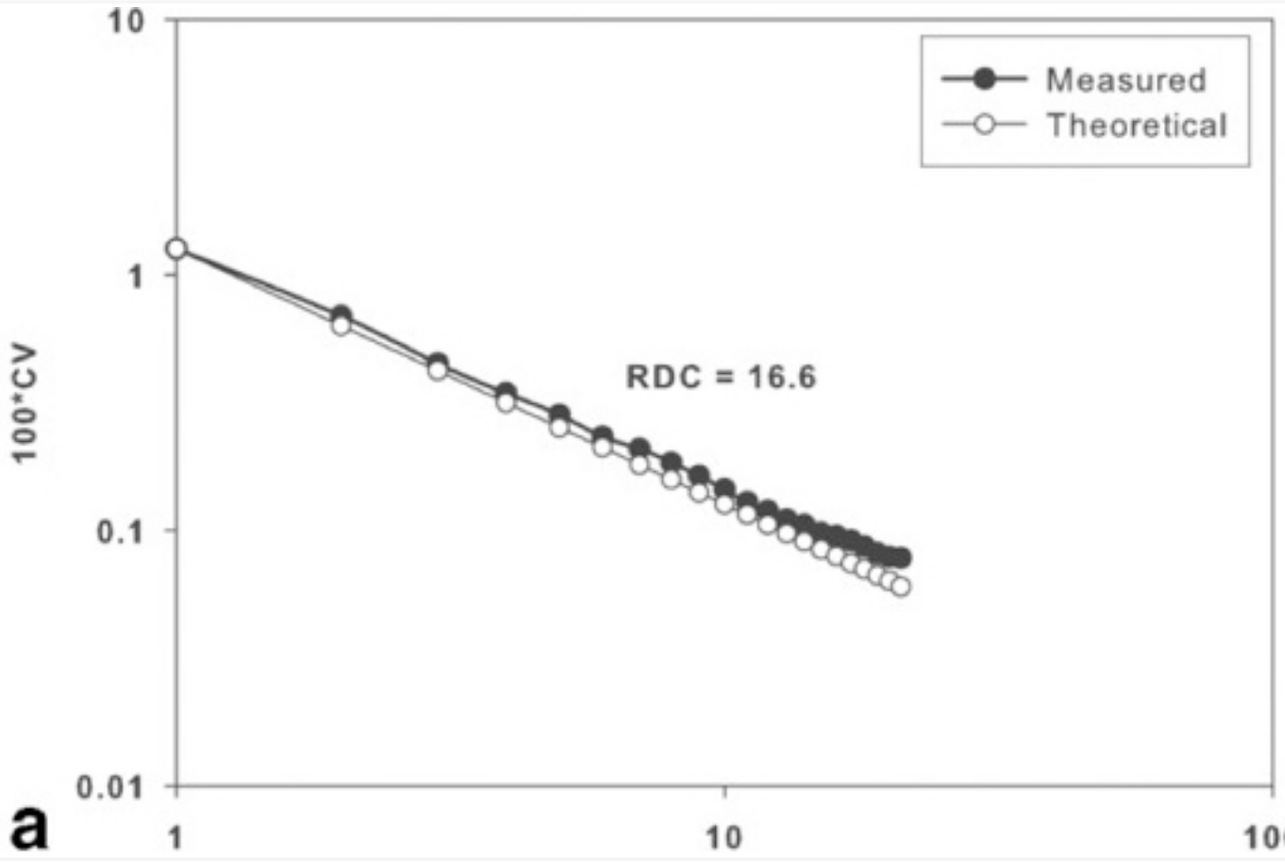


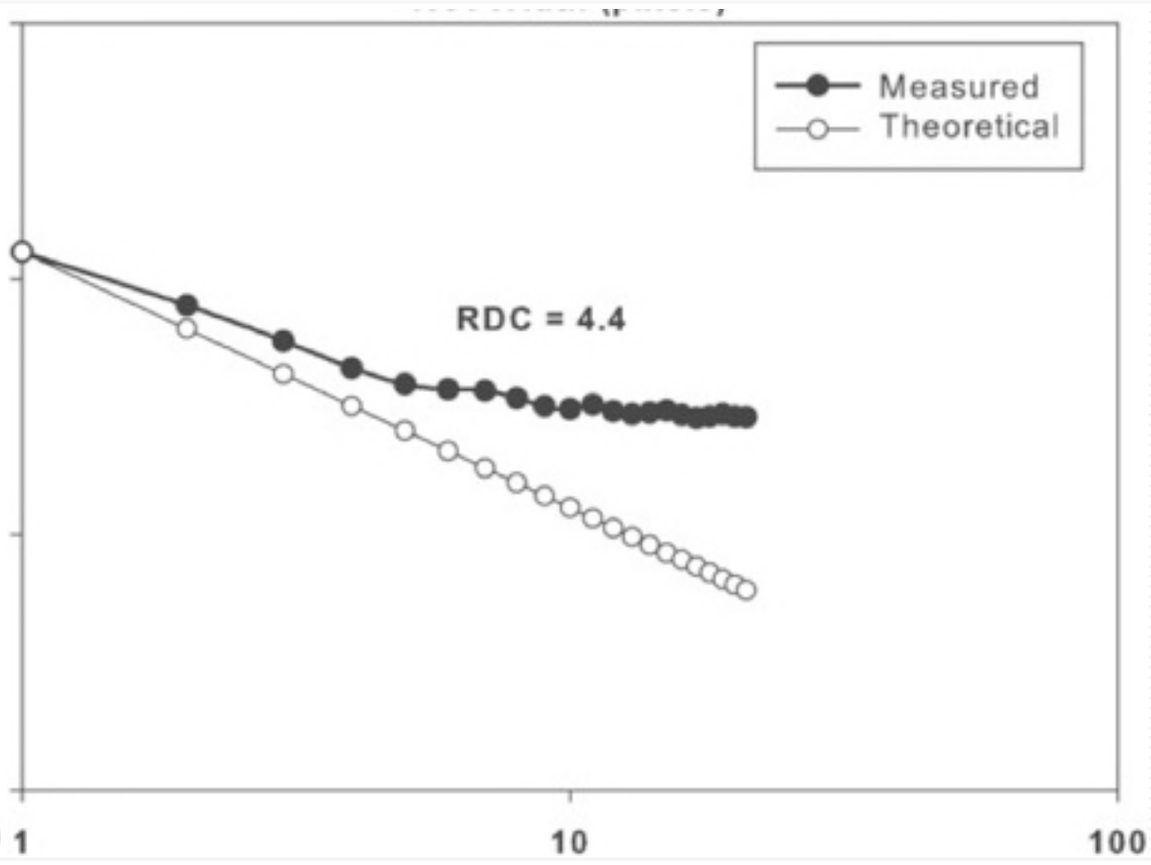
Deviations from the Theoretical Curve are Evidence of Correlated Noise

RDC (*Radius of Decorrelation*) is a Single Point Quantification of the Weisskoff Plot

Weisskoff Plot

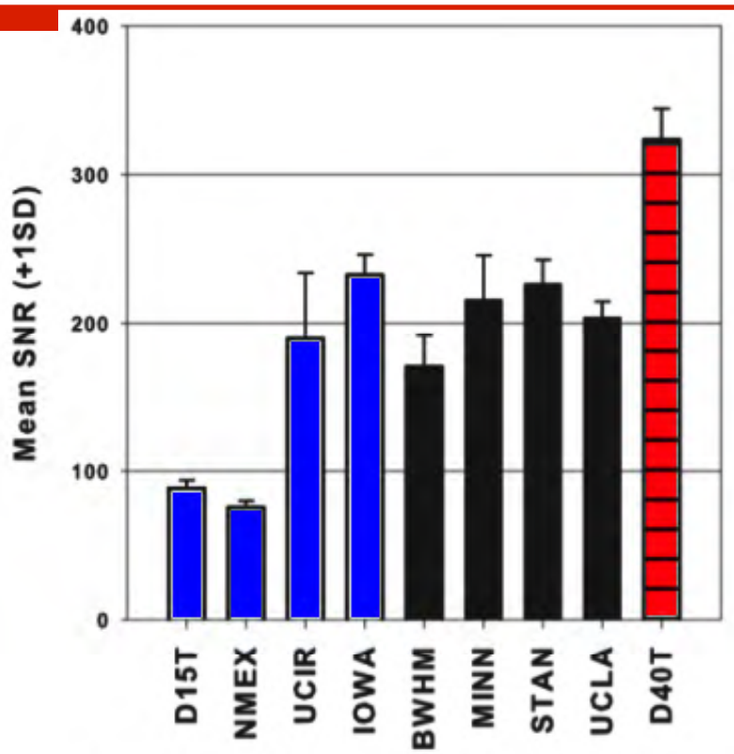
Weisskoff R. Magn Reson Med 36:643

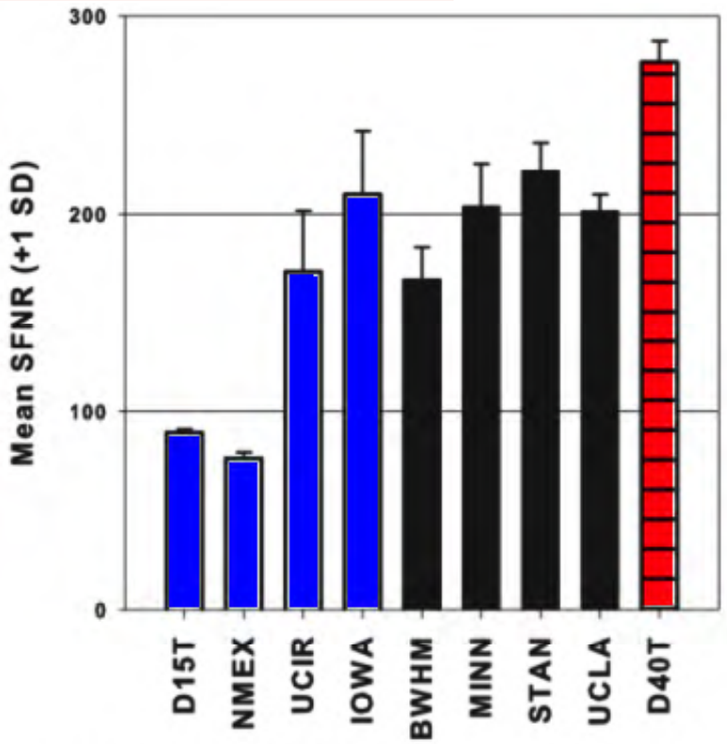


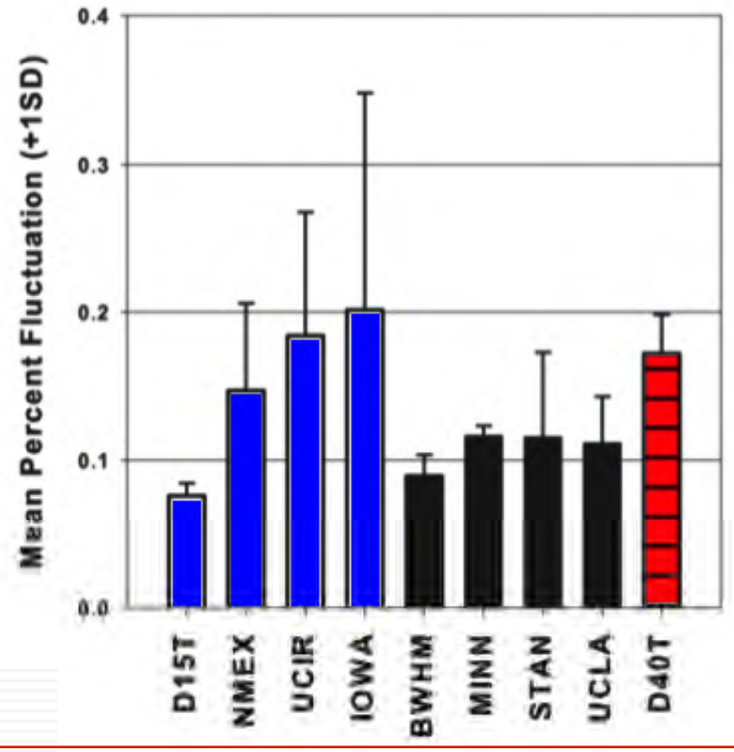


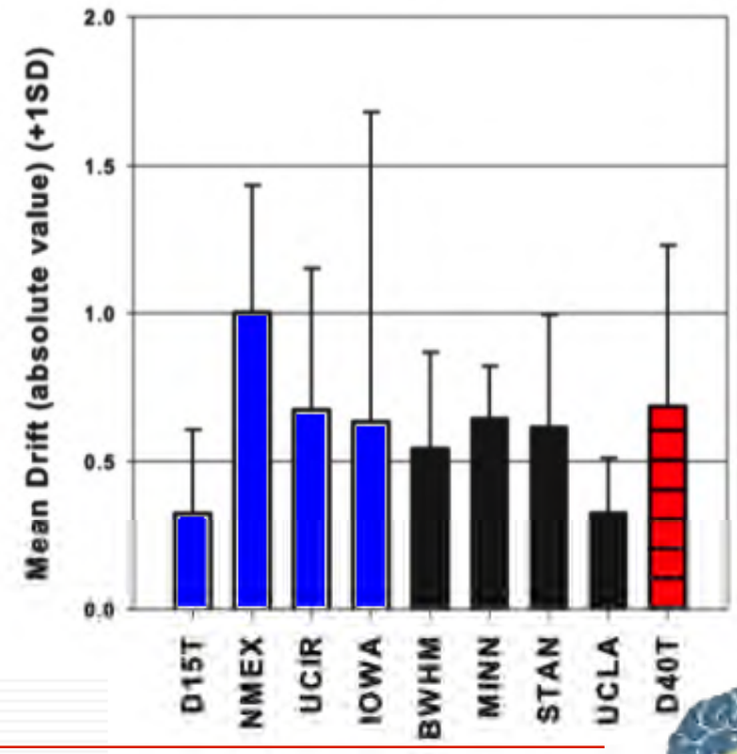
Friedman and Glover, JMRI 23:827

Scanner Comparisons



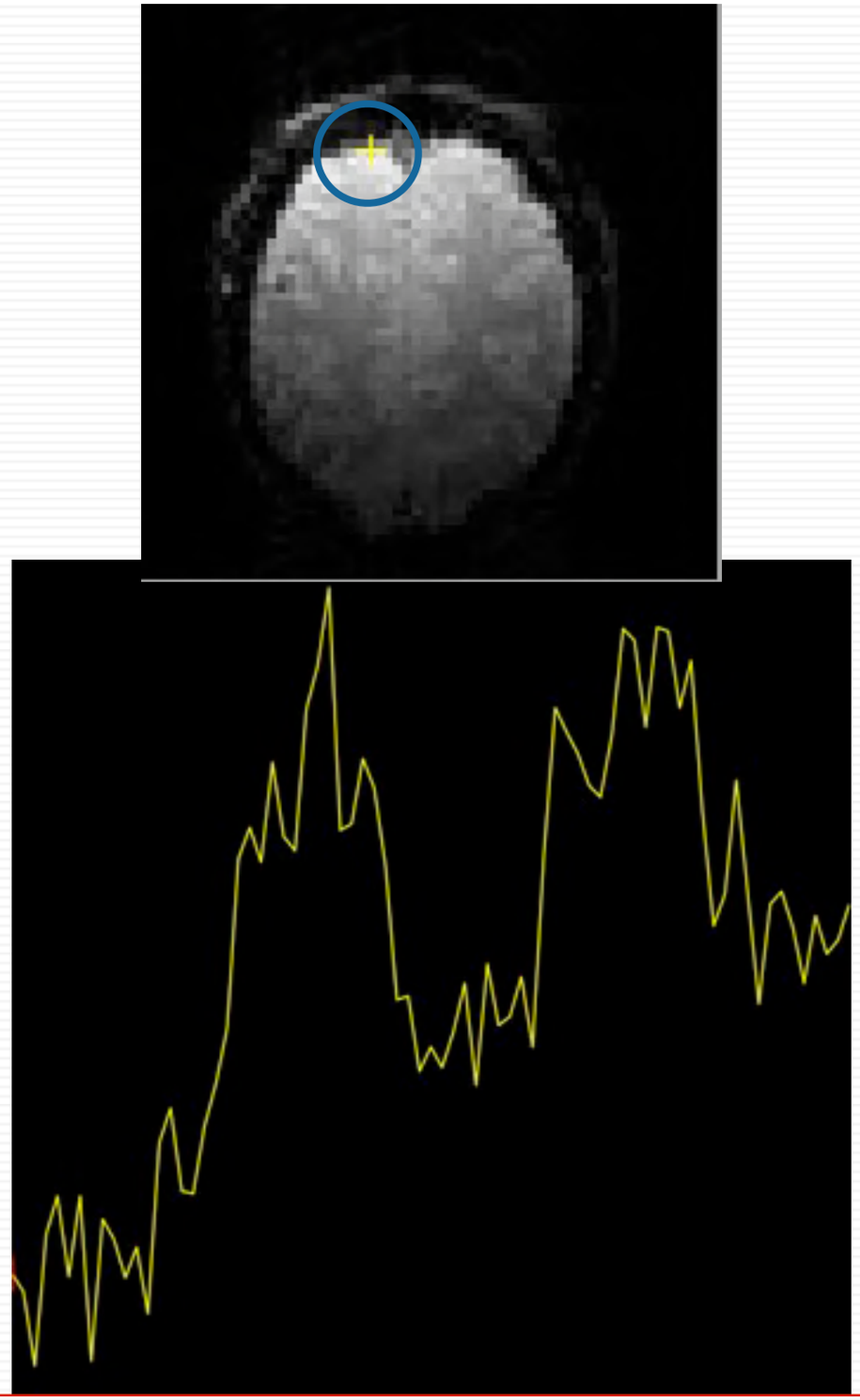


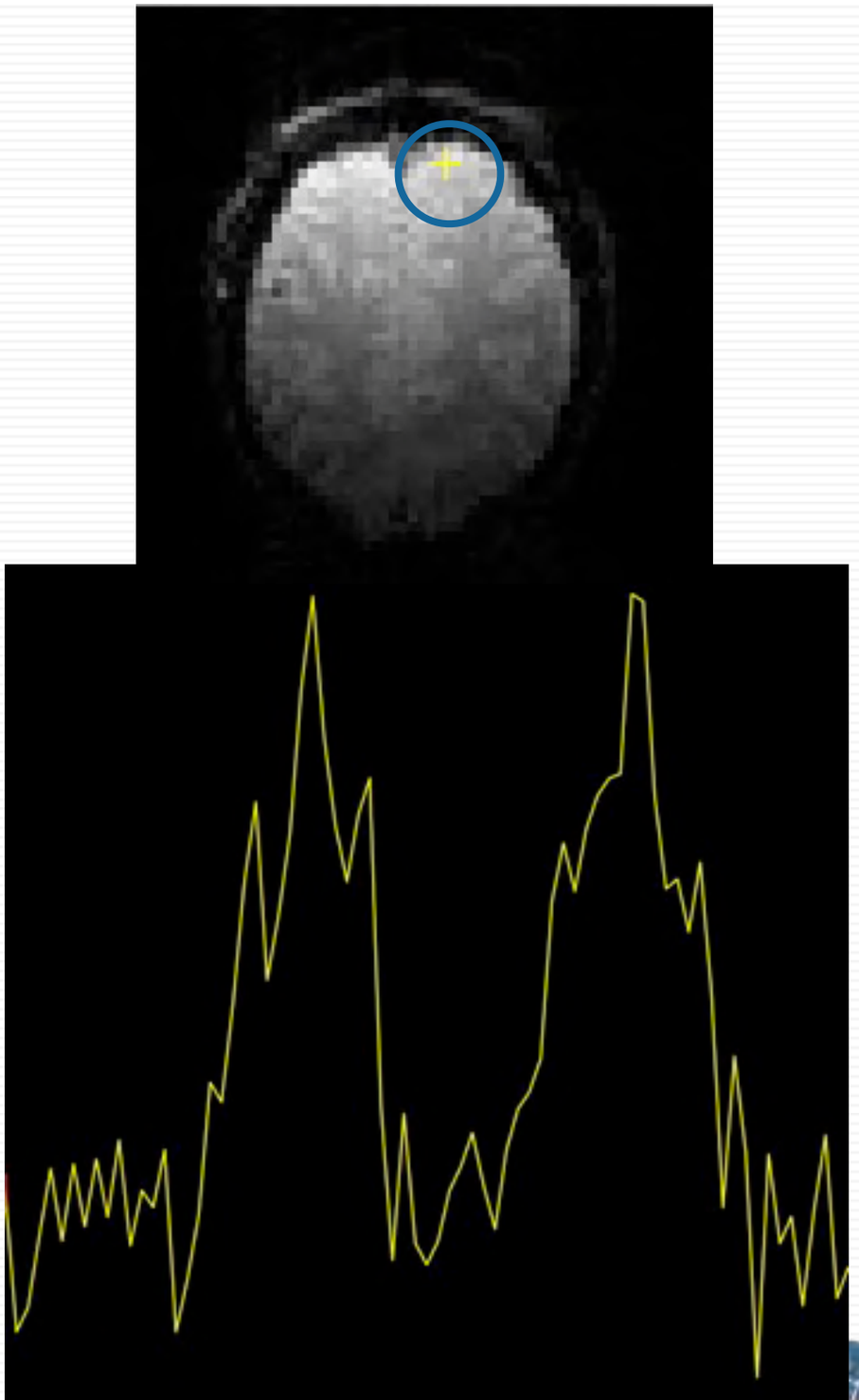




Friedman and Glover, JMRI 23:827

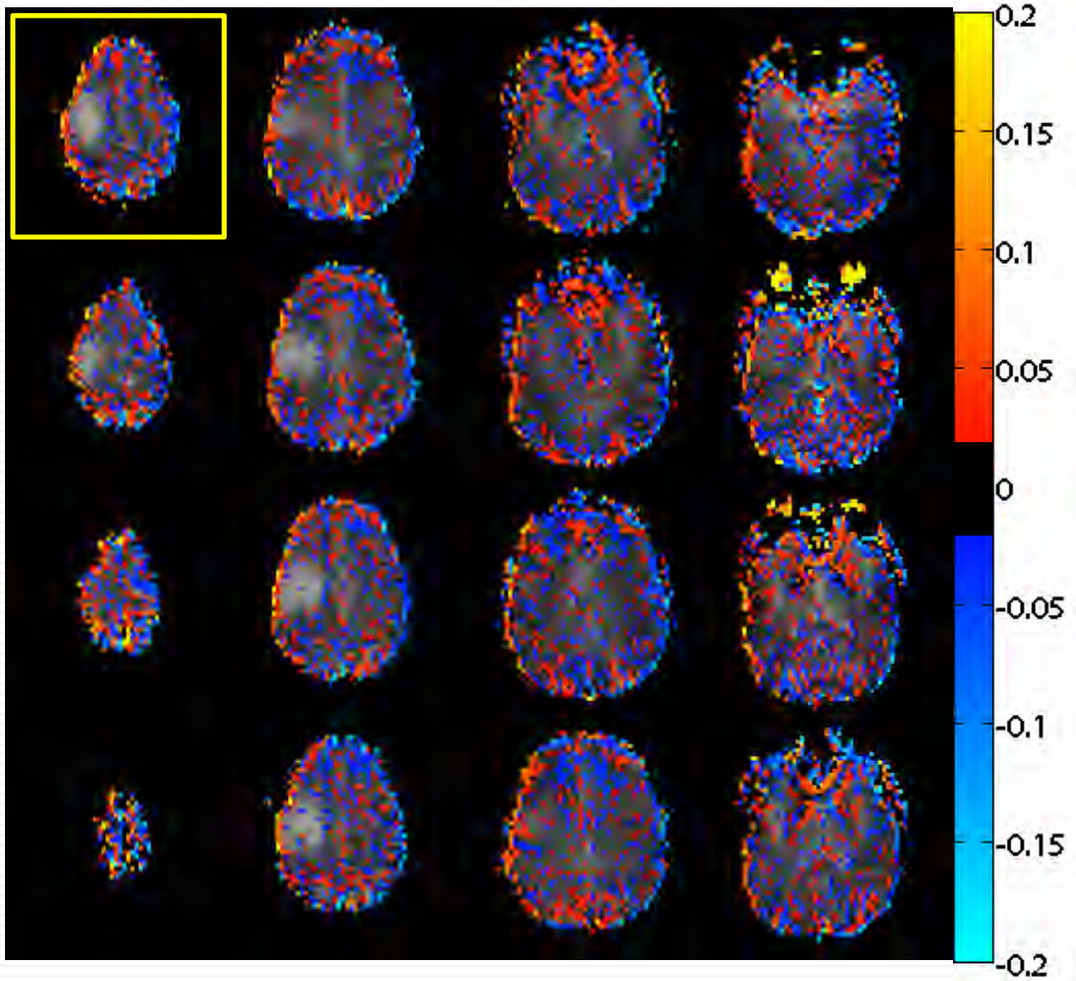
"Drift"

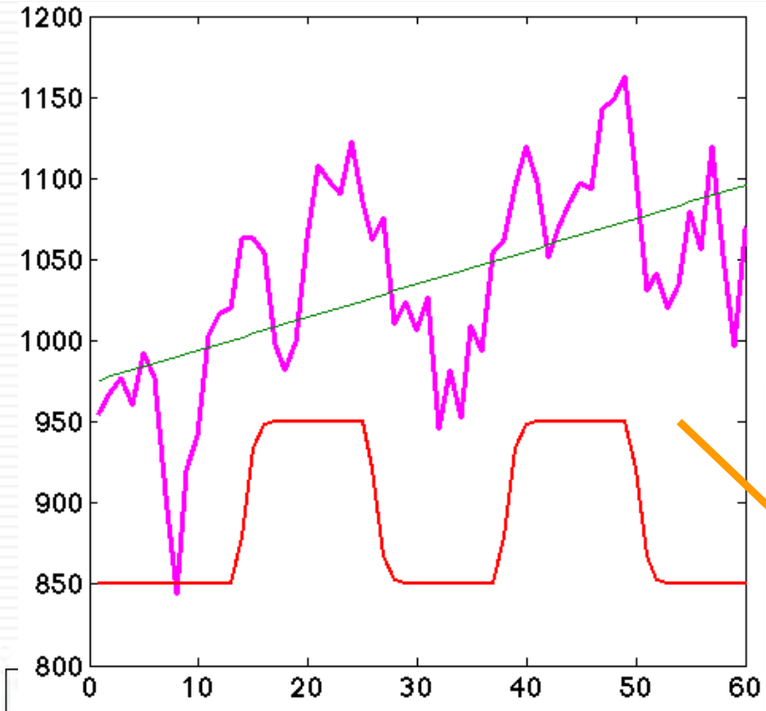


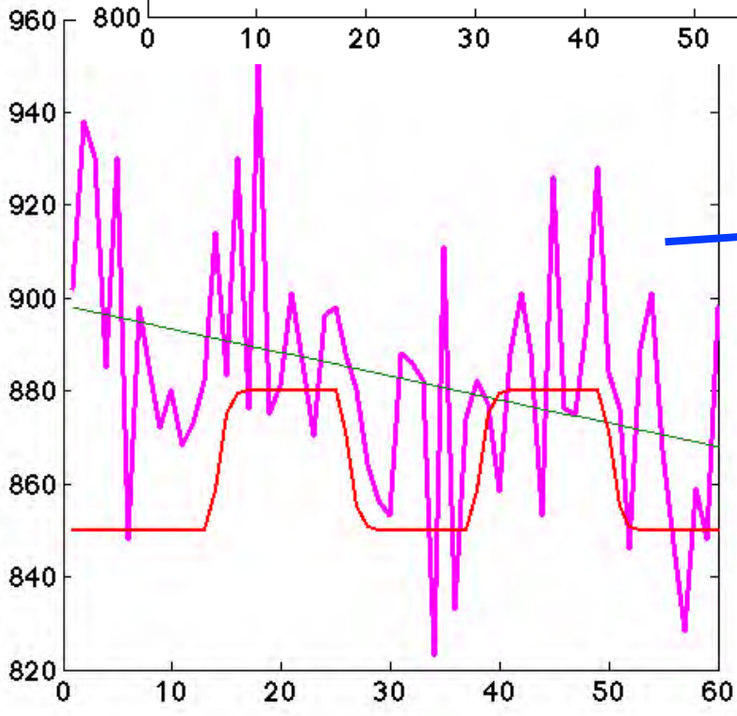


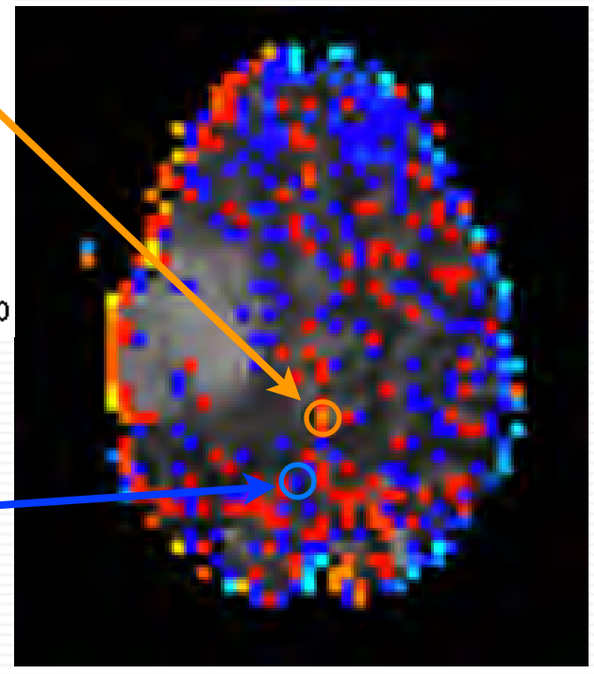
Instrument Variation

- 2. The mystery of scanner drift.

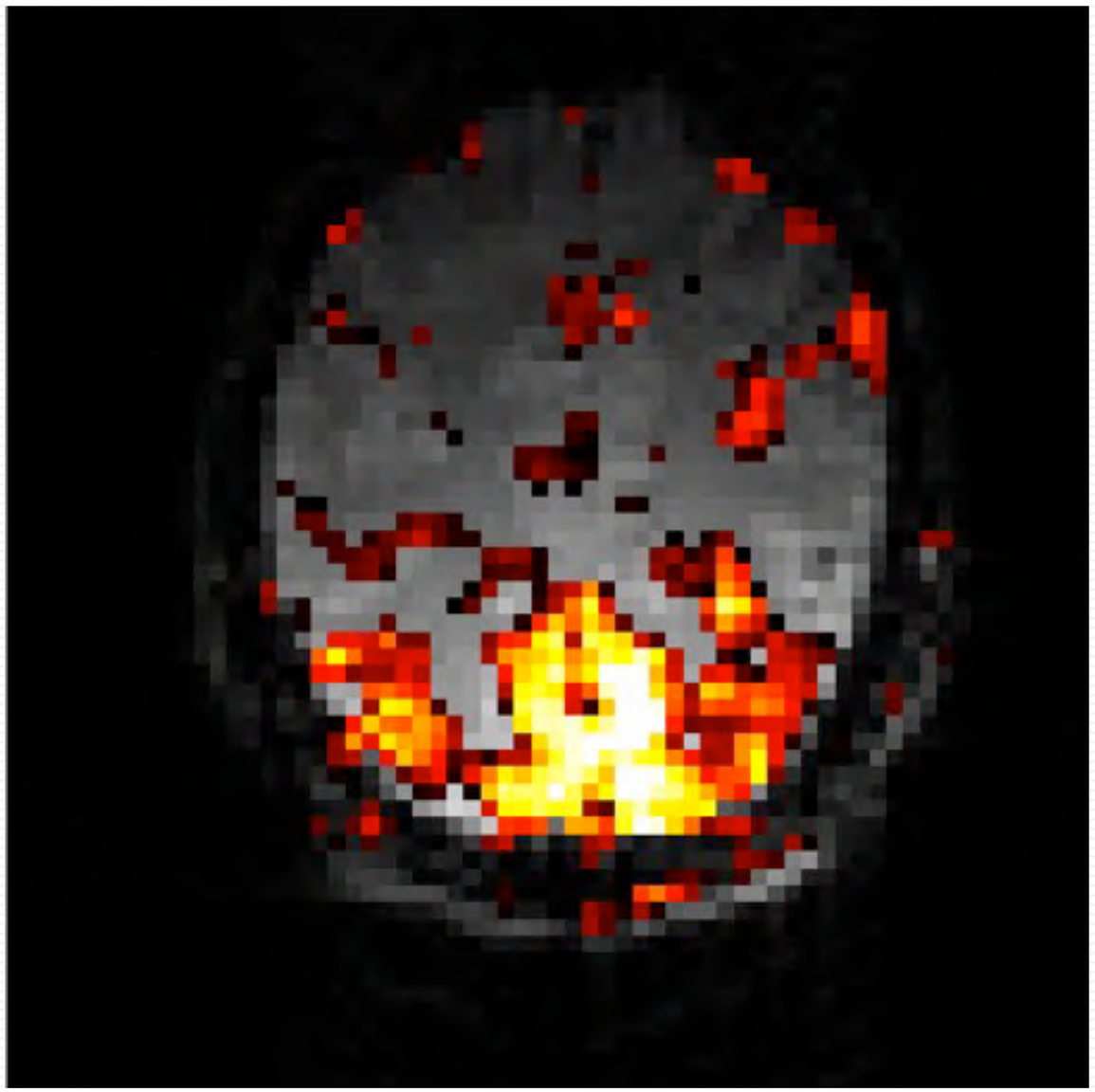


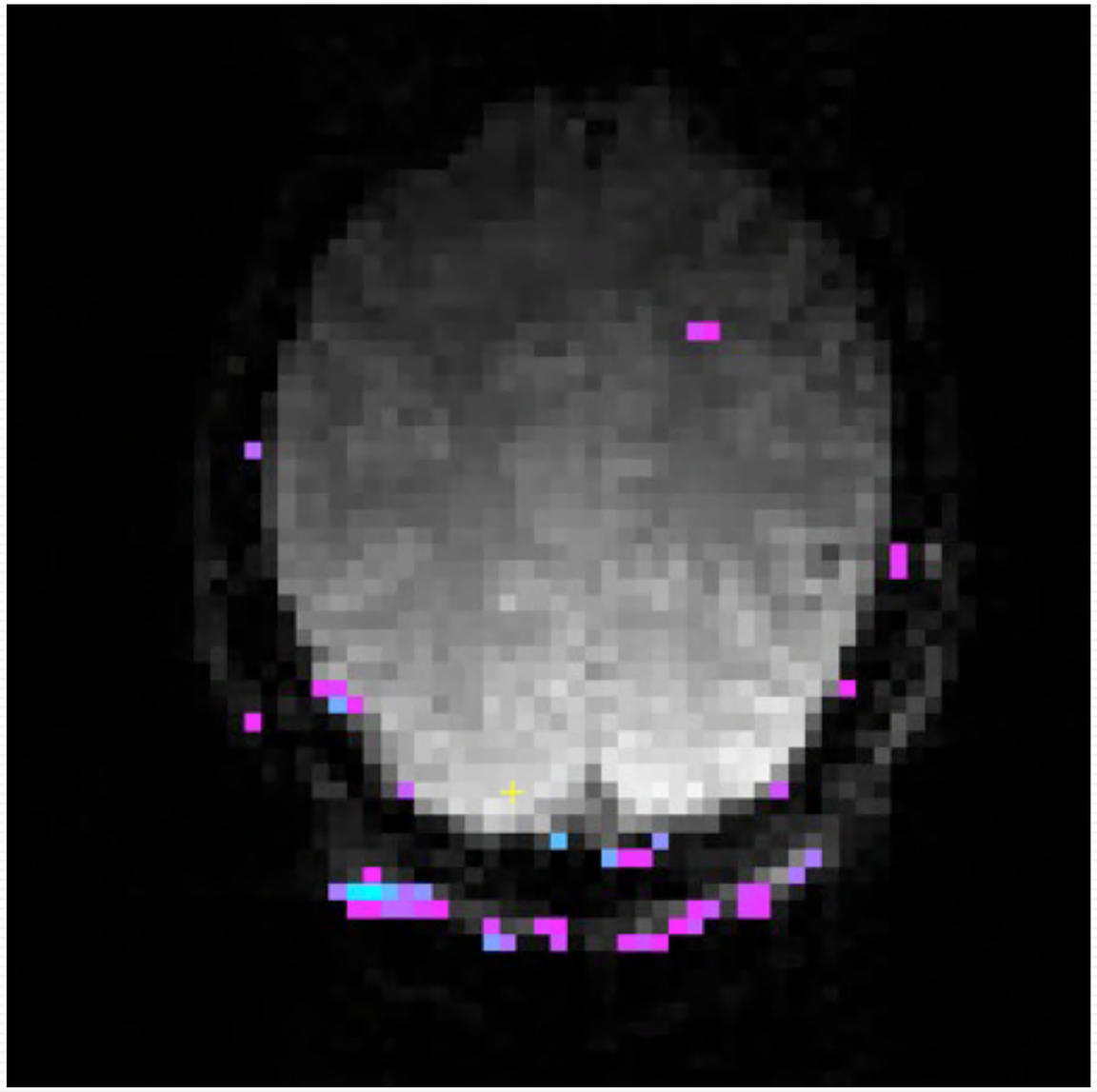




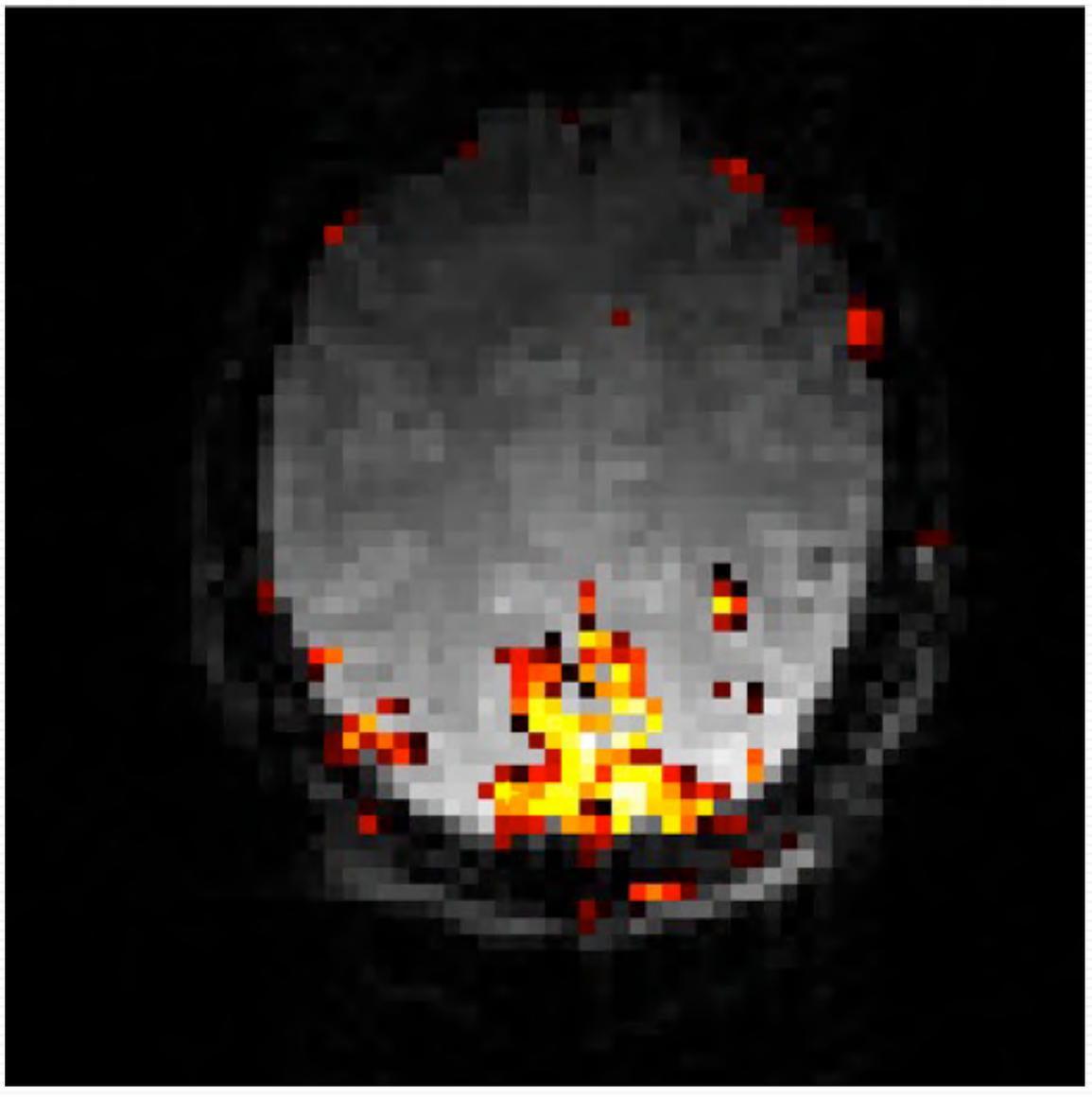


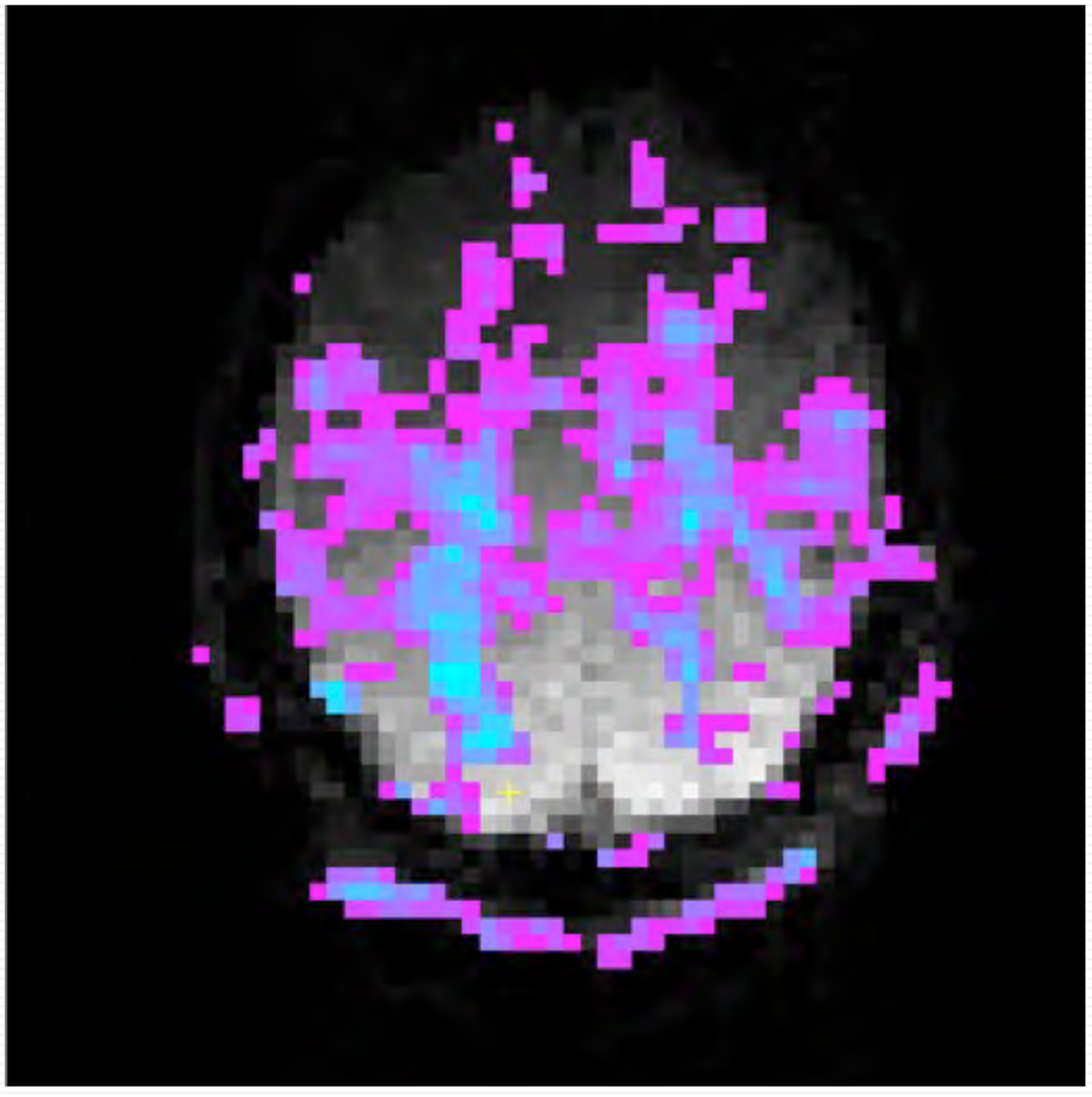
Global Mean Scaling - OFF





Global Mean Scaling - ON



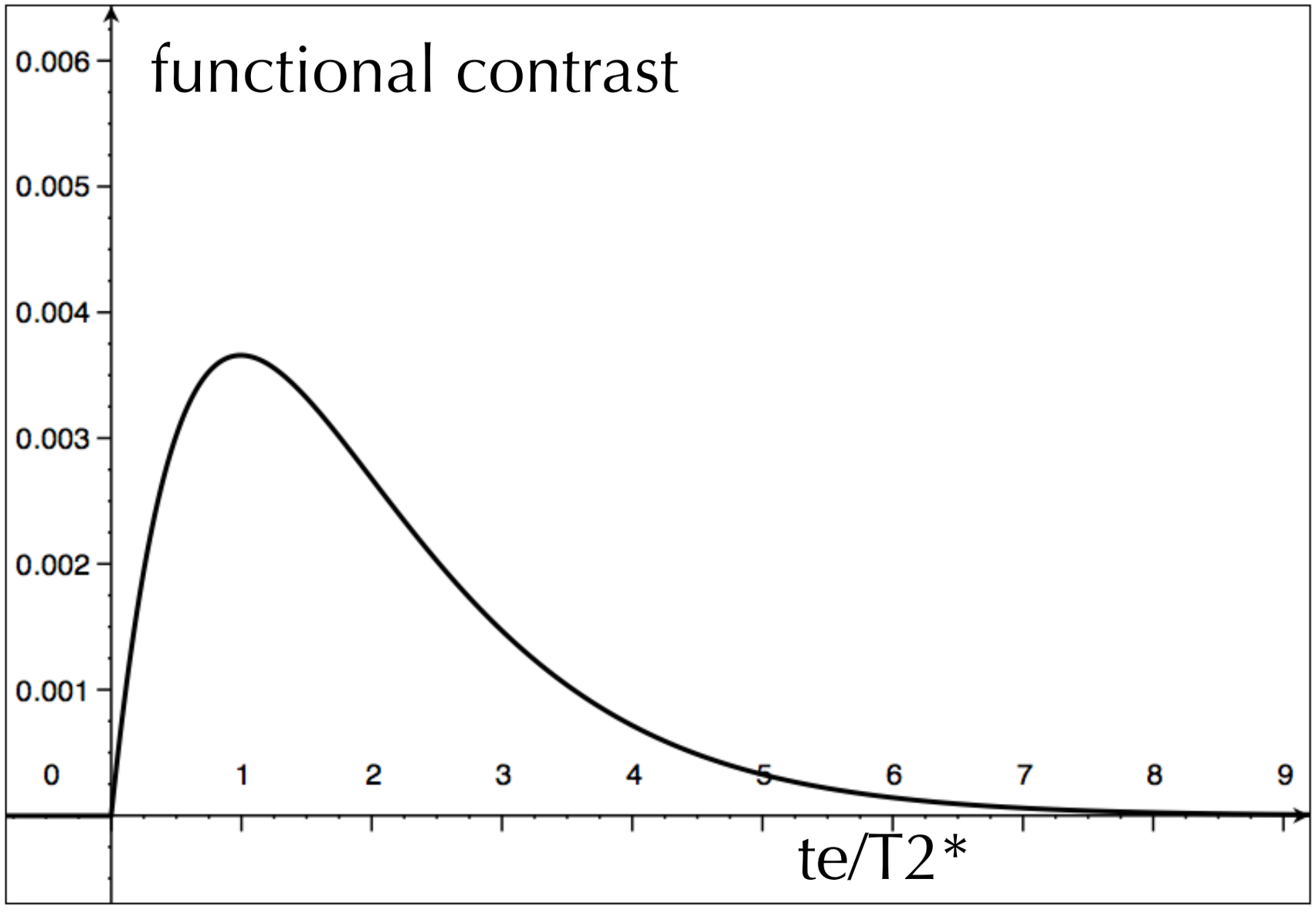


Thermal Noise

- Noise Distribution in MRI is **Rician**
  - $Signal = \sqrt{(\Re + \sigma_1)^2 + (\Im + \sigma_2)^2}$
- Background should be **Rayleigh** Noise $(\Re = \Im = 0)$
  - Expected **μ**: $\sigma\sqrt{\frac{\pi}{2}}$ expected variance: $\frac{4-\pi}{2}\sigma^2$
  - $\left\langle \frac{\mu}{\sqrt{\text{variance}}} \right\rangle = \frac{\sqrt{\pi}}{\sqrt{4-\mu}}$
- Deviations from This Model Imply Coherent Artifacts

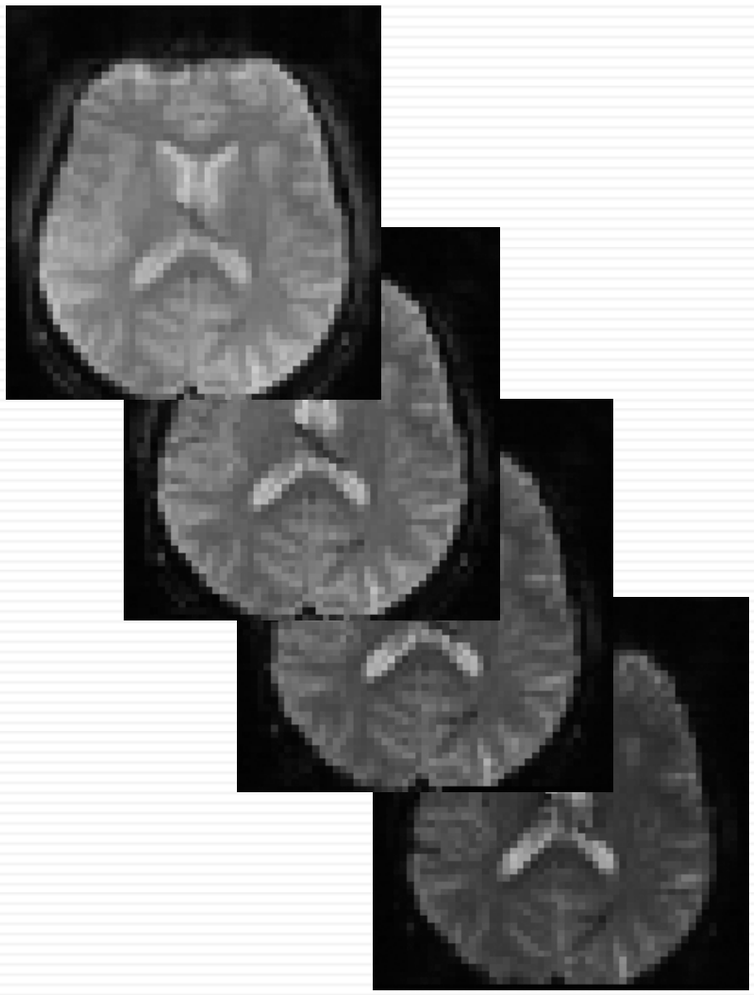
Parameter Optimization

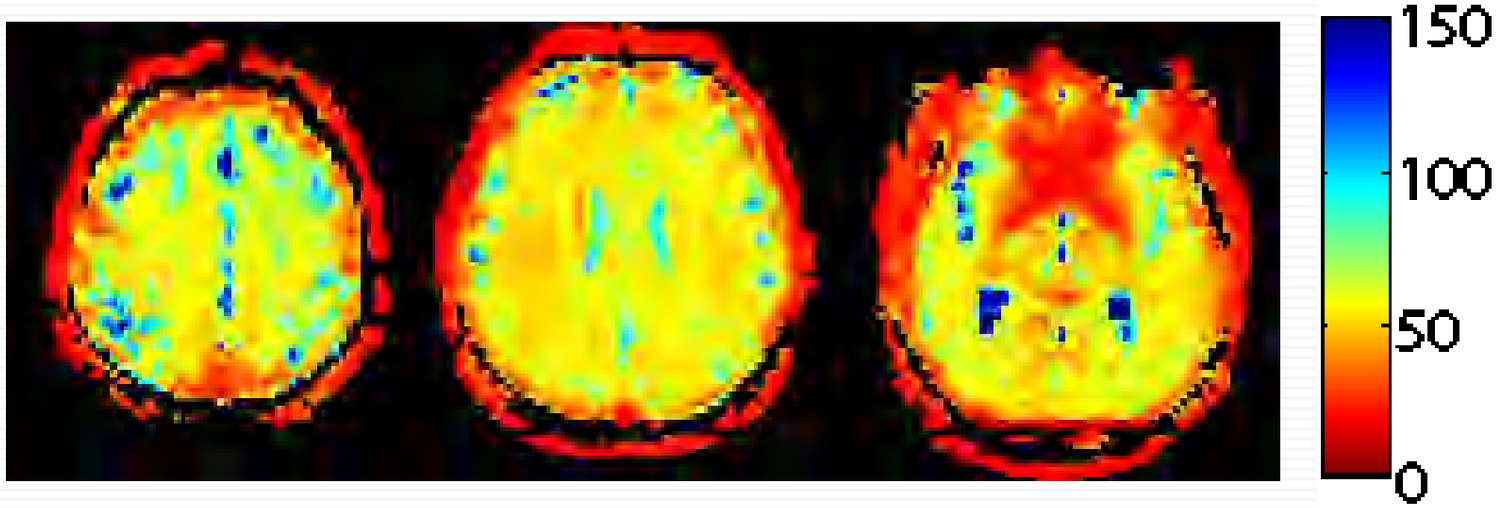
- Best BOLD contrast when te=T2*



Parameter Optimization

- Best BOLD contrast when te=T2*
  - $$\frac{d\ln(Signal)}{d(te)} = \frac{-1}{T2*}$$

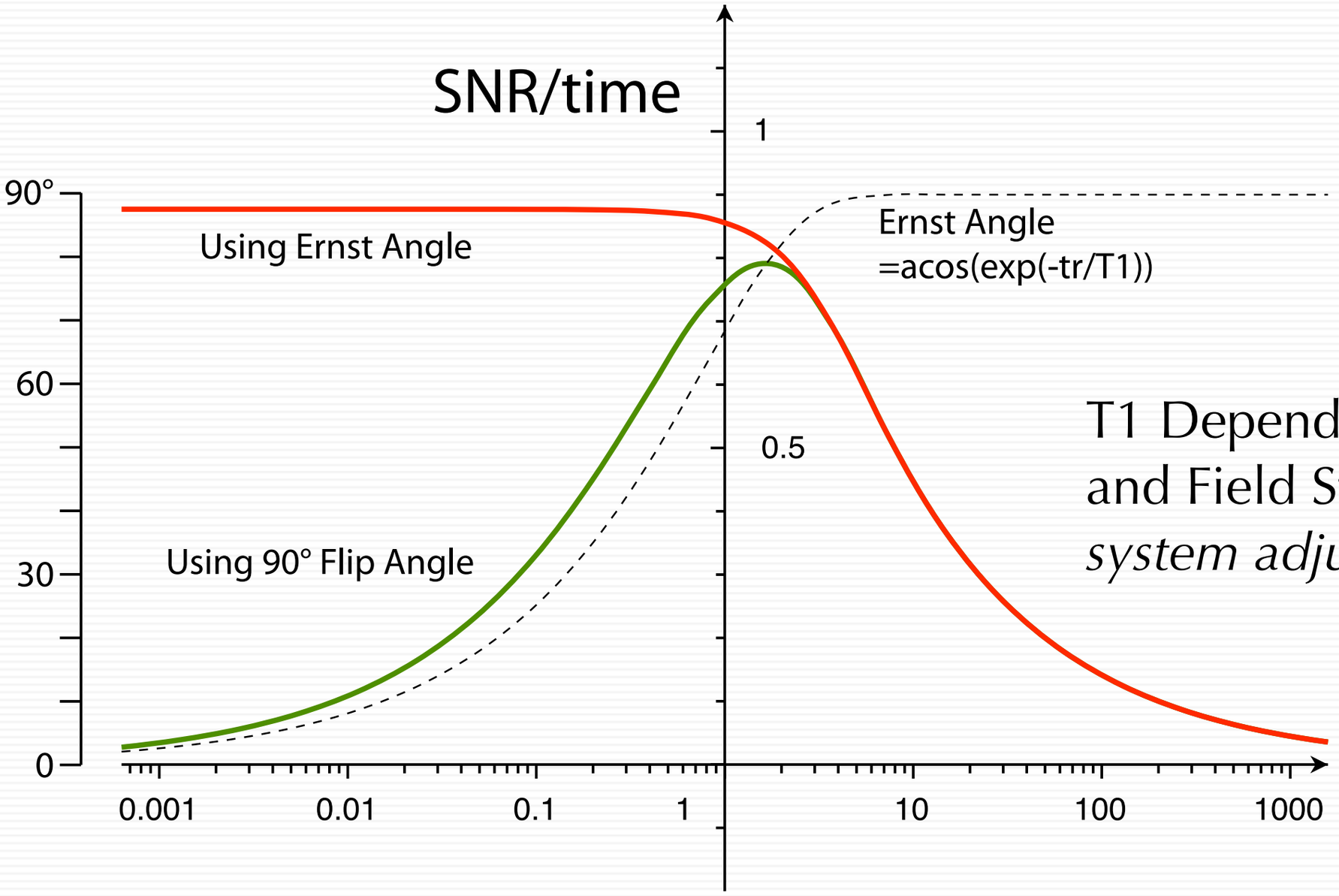




T2* is both an *Instrument* Parameter and a *Physiological* Parameter

Parameter Optimization - Flip Angle

- SNR per unit time is a *strong* function of tr/T1



T1 Depends on Tissue Type and Field Strength (*not on system adjustments*)

Other Parameters…

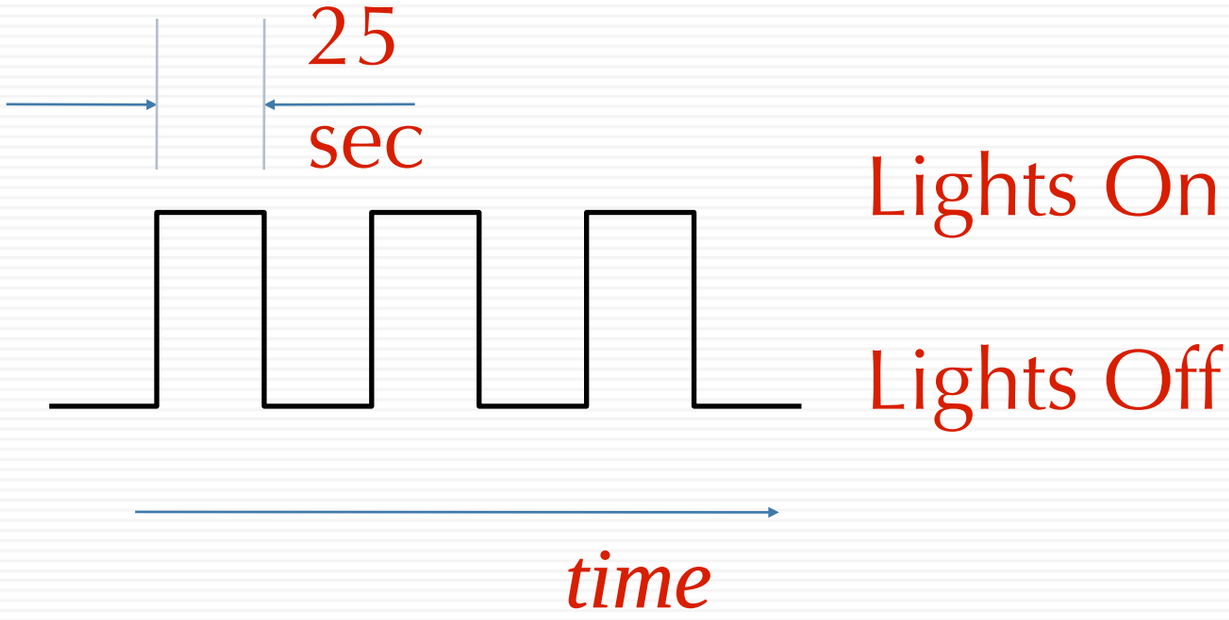
- Voxels Should be Large enough that:
  - Thermal Noise << Physiological Fluctuations
- For Best Signal with Arbitrary Slice Orientation:
  - Voxels Should be Isotropic
- With Gradient Echo Scans (most BOLD):
  - Signal Falls (much) more than linearly with Slice Thickness
- Do not Confuse Signal to Noise Ratio with Contrast to Noise Ratio!

Some Theoretical Considerations

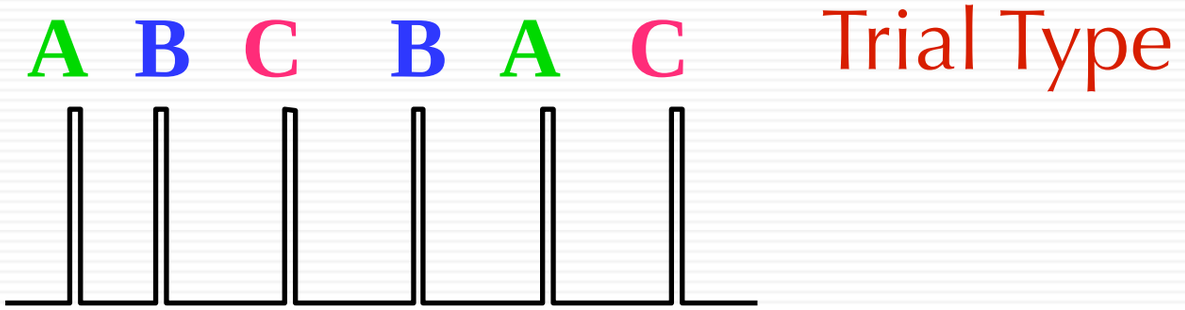
- Study Designs:
  - *Blocked*
  - *Single Trial*
- Predicting Responses
- Sources of Variance
- Resolution Limits:
  - *Temporal*
  - *Spatial*

Blocked vs. Single Trial

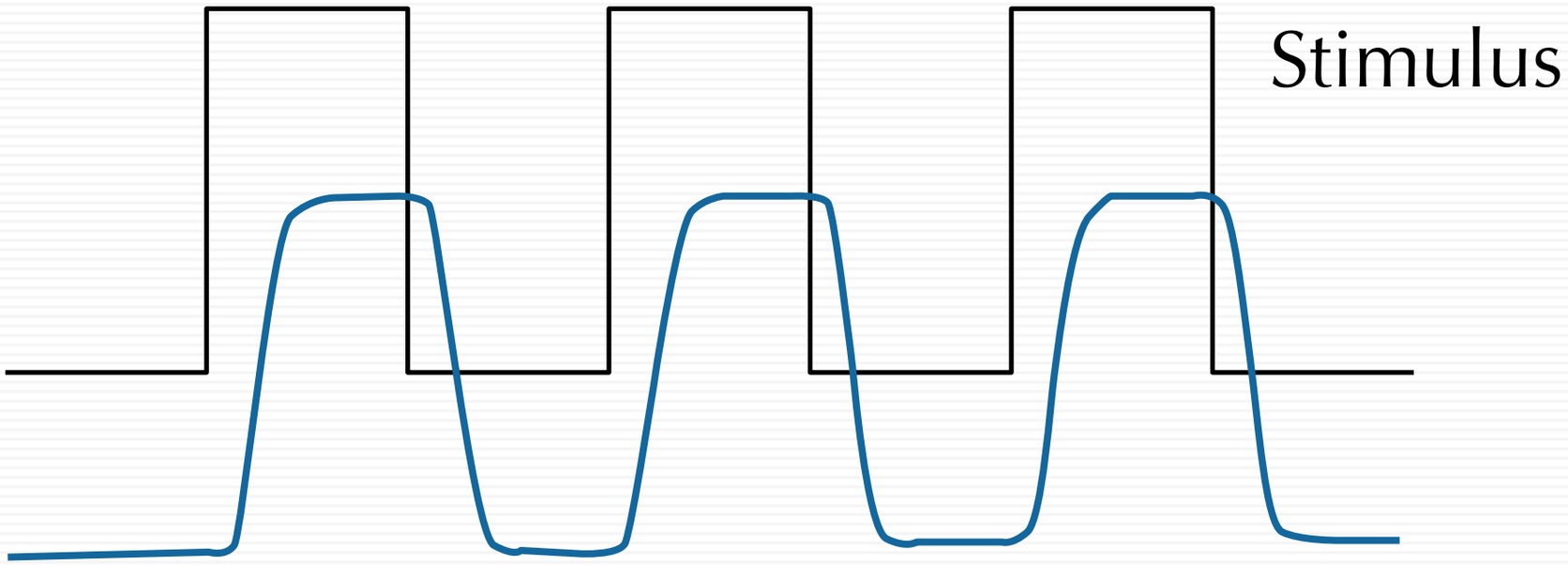
Typical Blocked Design



Typical Single Trial Design

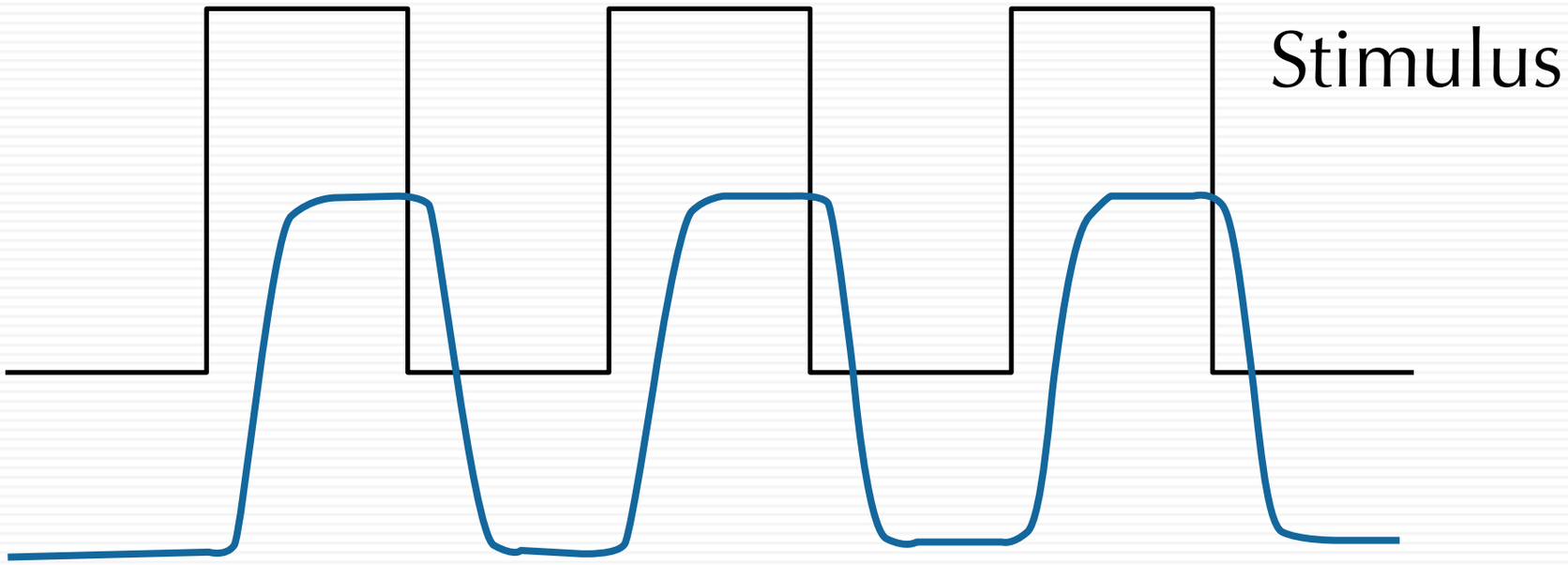


Blocked Experiments



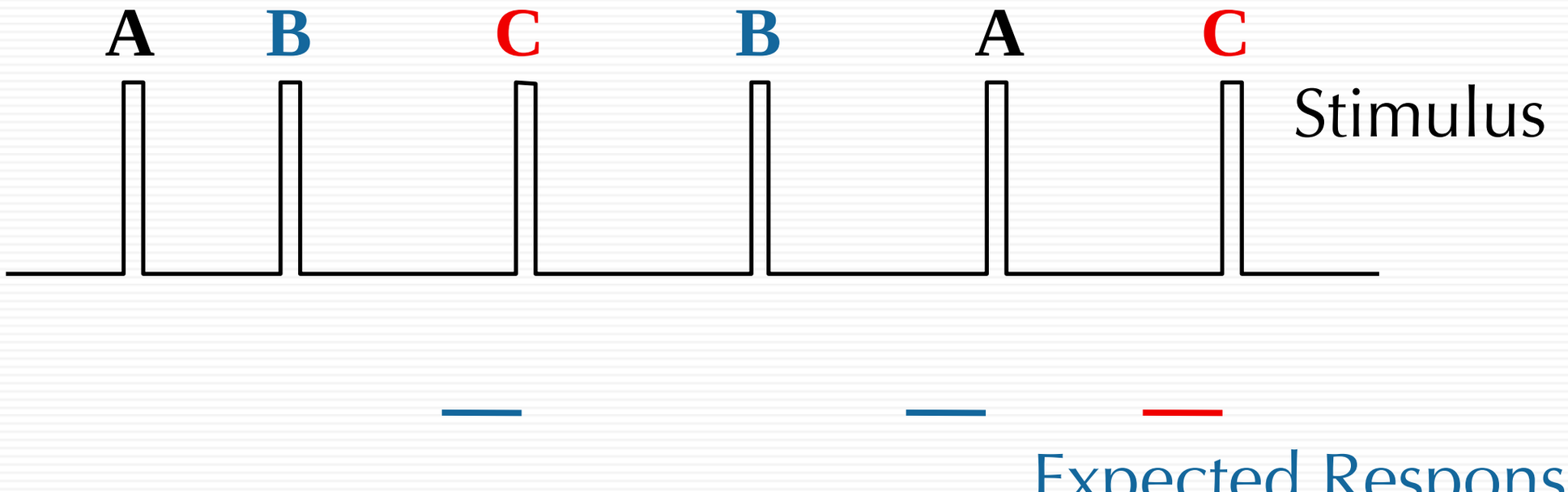
Expected Response

Blocked Experiments

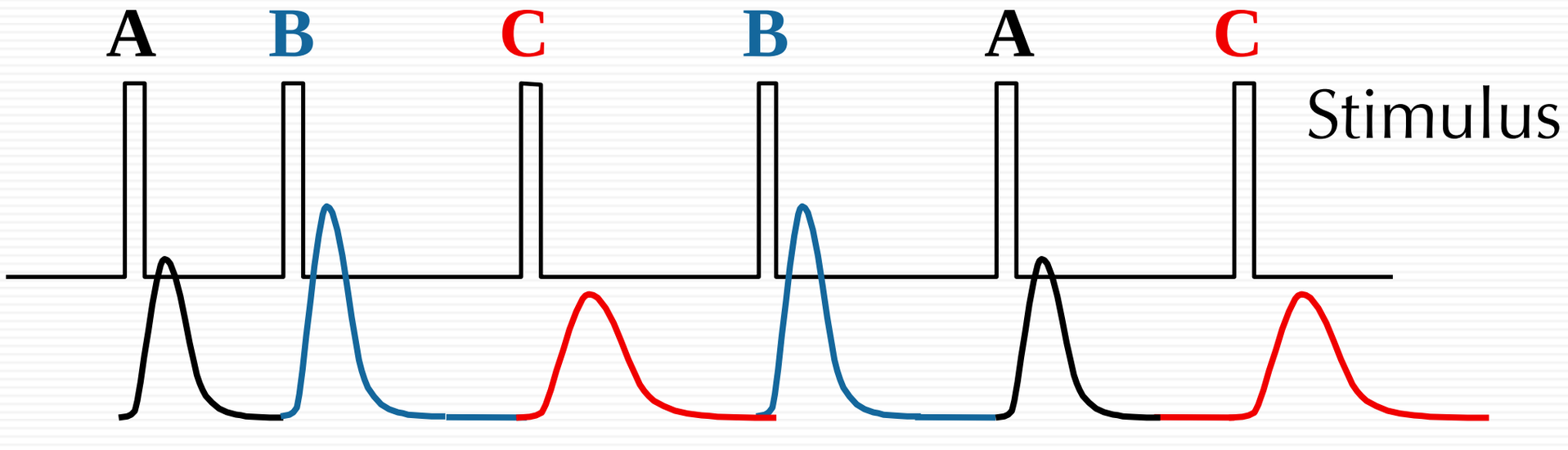


- Responses are Fairly Large
- Data are Easy to Analyze
- With Long Blocks, Time course can be Ignored
- All trials within a block are treated as Identical

Single Trial Designs



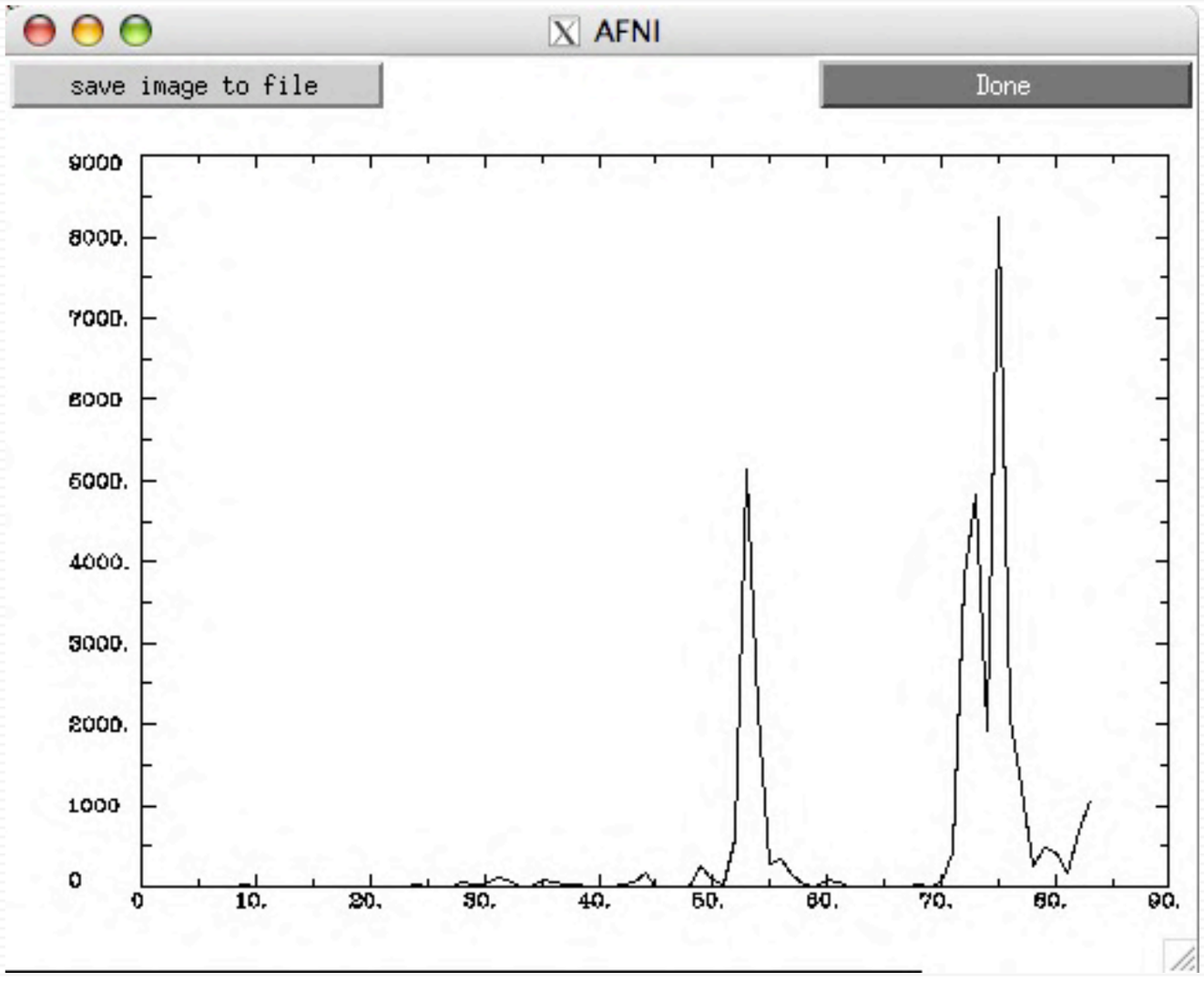
Single Trial Designs



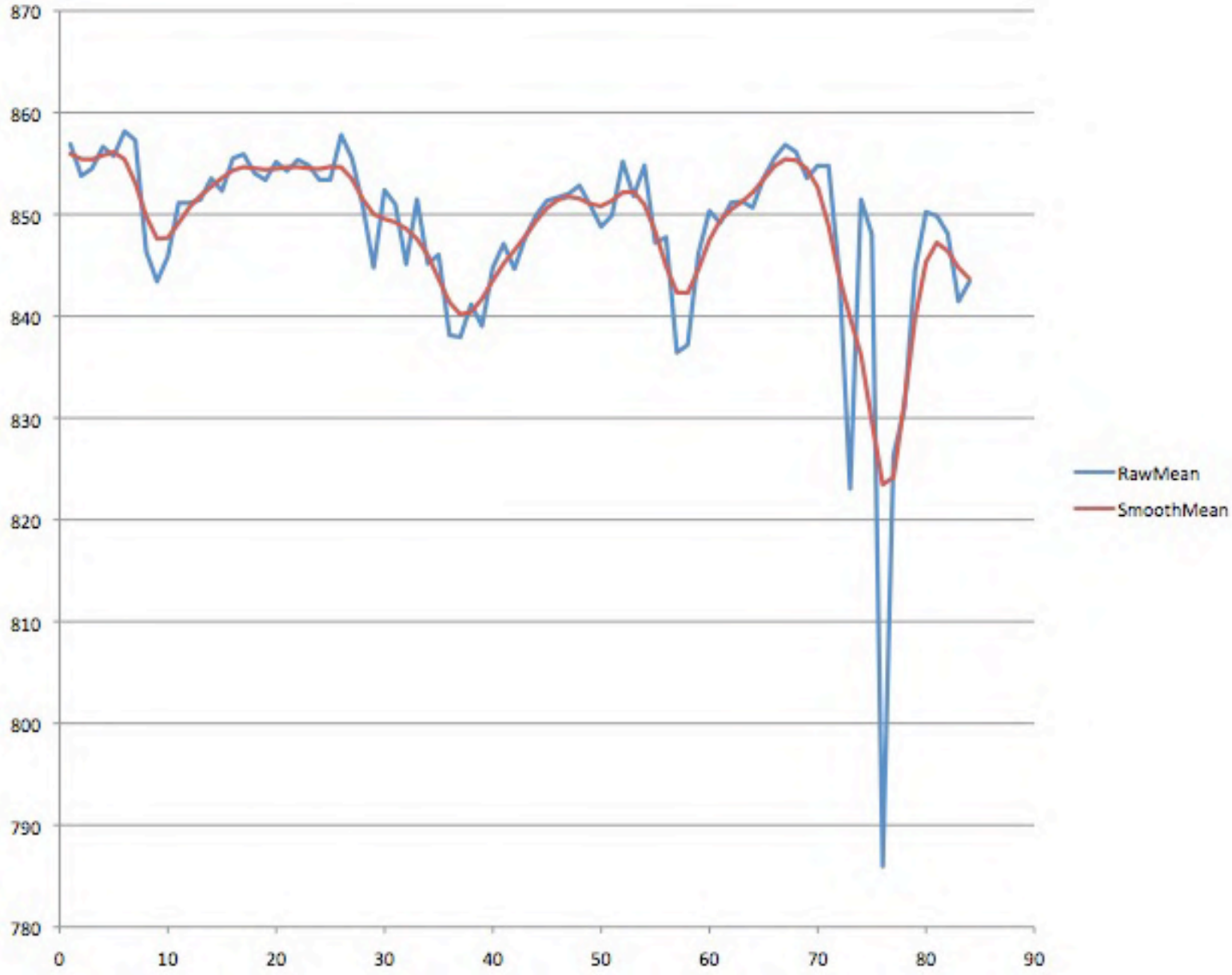
- Responses are small
- Useful contrast/noise is low
- Data are more Challenging to Analyze
- Exact Time course is Modeled or a Dependent Variable
- Suitable for Randomized Stimulus Designs

AFNI 3dToutcount

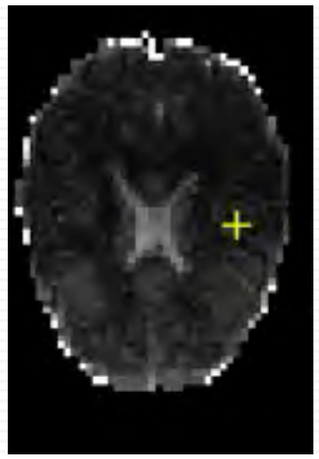
```
3dToutcount -automask KidAsImg.hdr | 1dplot -stdin
```

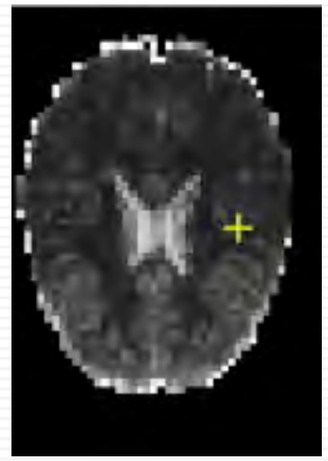


P2P



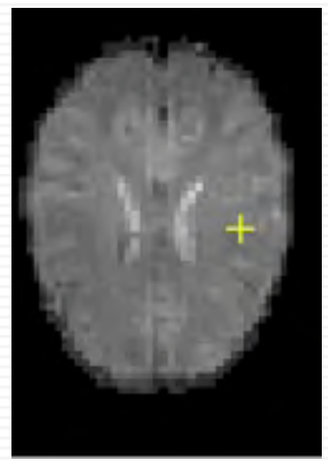
Peak-to peak





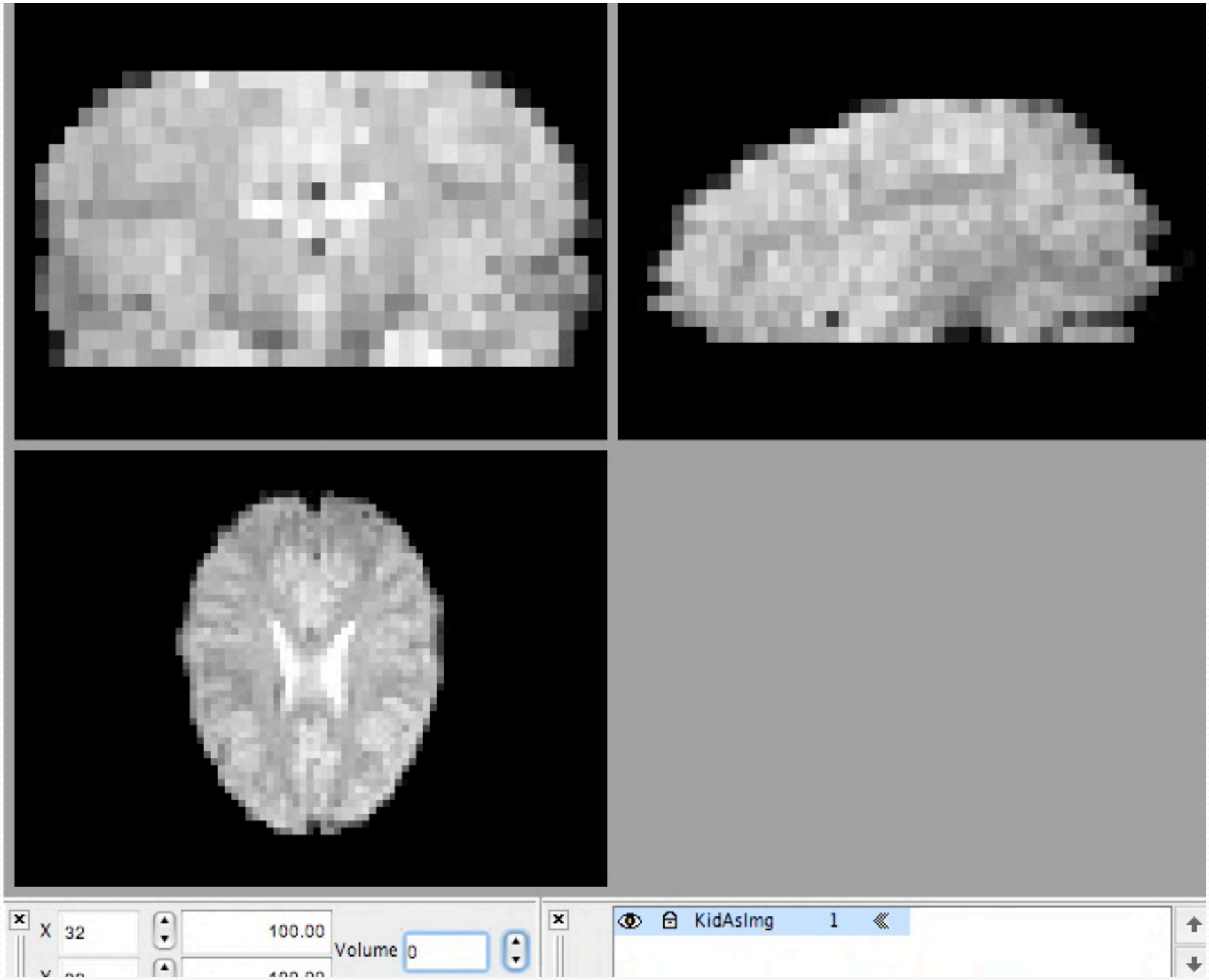
Std Dev

Max



Min

What really happened?



- HP filtering consequences
- FT of scanner drift
- Anti-alias at 2*tr
- linear regression for DCT gives fit values for the "nuisance" variables. Maps?
- Gaussian smoother is a low pass filter
- De-Meaning Figure