Welcome

NeuroImaging Training Program

Mark Cohen

UCLA Center for Cognitive Neuroscience

Departments of Psychiatry, Neurology, Radiology, Psychology, Biomedical Physics. Biomedical Engineering

UCLA Medical Center

Center for Cognitive Neuroscience

About the Neuroimaging Training Program

- Funded since 2006 by the National Institutes of Health
- Two components:
  - Graduate Fellowship Program - eligible to UCLA students
  - $100,000/year for the Summer Program
- Approximately 200 students

NIH RFA DA-06-011 and DA-11-006

Training in Neuroimaging: Integrating First Principles and Applications (T90)

Program Objective

This funding opportunity will enable the development of novel, interdisciplinary training programs that integrate comprehensive training in basic neuroscience, the physical and biological bases of neuroimaging, the technologies of *in vivo* neuroimaging, and the application of these technologies to understanding questions in neuroscience across the life span...

The goal of these training programs is to train the next generation of neuroimaging researchers who understand the underlying principles and the technologies of neuroimaging as well as their application to experimental questions in neuroscience. To realize this goal, it is imperative to recruit and expose students early in their careers to the ways in which their interests can be applied to questions in neuroscience through the mathematical, physical, and chemical principles of neuroimaging. Training programs are required to interface trainees from the quantitative, engineering, and physical/chemical sciences with trainees from biomedical/biological disciplines in the same integrated training program.

MULTIMODAL NEUROIMAGING TRAINING PROGRAM - U. Pittsburgh, Seong-Gi Kim
ADVANCED MULTIMODAL NEUROIMAGING TRAINING GRANT - Harvard, Bruce Rosen
COMPREHENSIVE TRAINING IN NEUROIMAGING FUNDAMENTALS AND APPLICATIONS - UC Los Angeles, Mark Cohen

You

- 36 Students + 6 NITP Fellows
- 7 Countries
- 4.4 First Author Papers
- 5.9 Co-author Papers

Procedures

- All Sessions are Webcast - Students are NOT filmed
  - www.brainmapping.org/NITP/Summer2011.php
- We try to get Speakers Slides (not always successful)
- Breakfast Provided
- Lunch noon-1 on your own
- Afternoon Labs in Faculty Center
- Laptops available to share
- Course Software Available by Request
- Wireless provided in classroom

Schedule

	7/11	7/12	7/13	7/14	7/15
8:30	**Intro and Overview** Cohen, Bookheimer	**Psychophysics for fMRI** Lenartowicz	**Resting State and Population Studies** Greicius	**fMRI data group analysis** Mumford	**Principles and Pitfalls of Machine Learning** Müller
9:30	**MRI acquisition I** Cohen	**Expermimenal design I** Bookheimer	**Resting State ICA and Advanced Analytic Methods** Greicius	**Setting up Models for fMRI** Mumford	**BCI, fMRI and tkCCA** Müller
10:30	Break				
11:00	**MRI acquisition II** Cohen	**Exp design II** Bookheimer	**fMRI Single Subject Analysis** II Monti	**Setting up Models for fMRI (*workshop*)** Mumford	**Non-parametric Imaging Statistics** Lindquist
Noon	Lunch				
1:15	**Hemodynamics** Buxton	**fMRI preprocessing** Monti	**fMRI artifacts/quality control** Cohen	**Multiple comparisons** Lindquist	**Bayesian Statistics** Lindquist
2:15	**Relating neurophysiology and imaging signals** Buxton	**Statistical Inference** Monti	**FSL Refresher/Advance** - Bramen, Brown	**Safety issues in neuroimaging** Cohen	**Functional Connectivity** Rissman
3:30	**Psychtoolbox** I - Lenartowicz	**Psychtoolbox** II- Lenartowicz	**Warping** - Burggren **Group breakouts - finalize study design**	**Group Project Development**	**Functional Connectivity Lab** Rissman Brown
Saturday and Sunday - Run Group-Designed Experiments at 3 Tesla					

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Schedule

	7/18	7/19	7/20	7/21	7/22
8:30	**Reporting Whole Brain Data; Pipeline** Van Horn	**Psychopathogy and Imaging Genetics** Cannon	**Diffusion Spectrum Imaging I** Wedeen	**ICA for Discovery and Inference Testing I** Beckmann	**Some thoughts on setting up a lab** Cohen
9:30	**Imaging-genetics** Glahn	**Real-time fMRI** LaConte	**Diffusion Spectrum Imaging II** Wedeen	**ICA for Discovery and Inference Testing II** Beckmann	**Group Presentations**
10:30	Break				
11:00	**Machine Learning** Pereira	**Group ICA of FMRI: Introduction and Review of Current Work** Calhoun	**Network Analysis I** Bassett	**New Methods of Connectiviy Analys** Laurienti	**Group Presentations**
Noon	Lunch				
1:15	**Machine learning for Abstract Feature Detection** Pereira	**An ICA Framework for Fusion of Multimodal Imaging and Genetic Data** Calhoun	**Network Analysis II** Bassett	**Greg Simpson** Network Dynamics	**Group Presentations**
2:15	**Machine Learning Lab** - Pereira, Douglas	**ICA with GIFT** Calhoun	**High Resolution Tractography: Diffusion Toolkit** Wedeen	**Group ICA Lab Christian Beckmann**	**Reception**
3:30	**Machine Learning Lab** - Pereira, Douglas	**ICA with GIFT** Calhoun	**High Resolution Tractography: TrackVis** Wedeen	**Group Project Analyses**	**Q&A Ask the Experts**

Evaluations

- Please link this page:

  http://www.brainmapping.org/NITP/Summer2011Evals.php

- Please remember to fill out your evaluations for each speaker and lecture.

- These evaluations are extremely important to us in keeping the course strong.

?

Magnetic Resonance Imaging

Mark Cohen

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NeuroImaging Training Program

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

|ˈmetəˌfôr; -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.

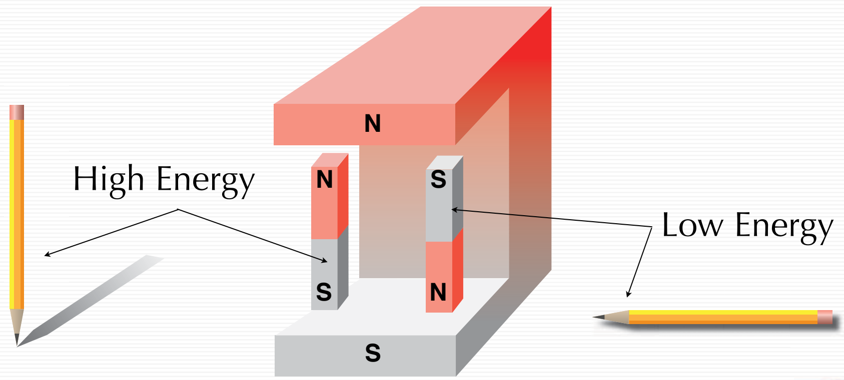
a thing regarded as representative or symbolic of something else, esp. something abstract *:*

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

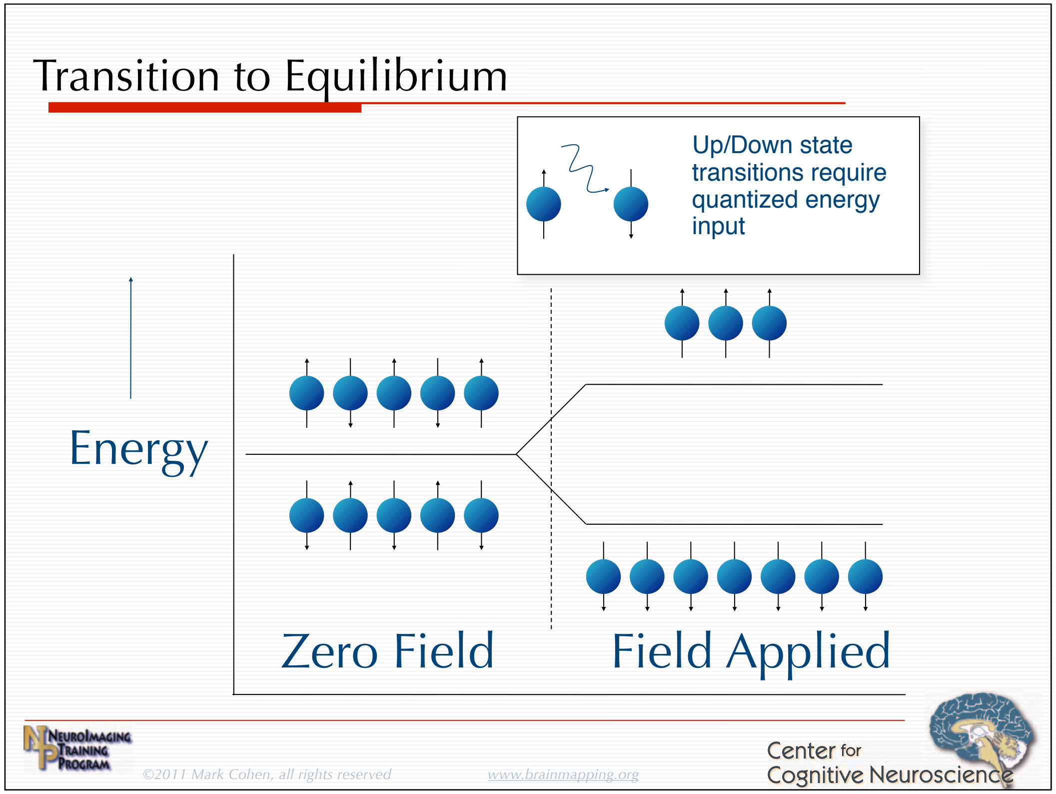
2. Little white lie

Spin States

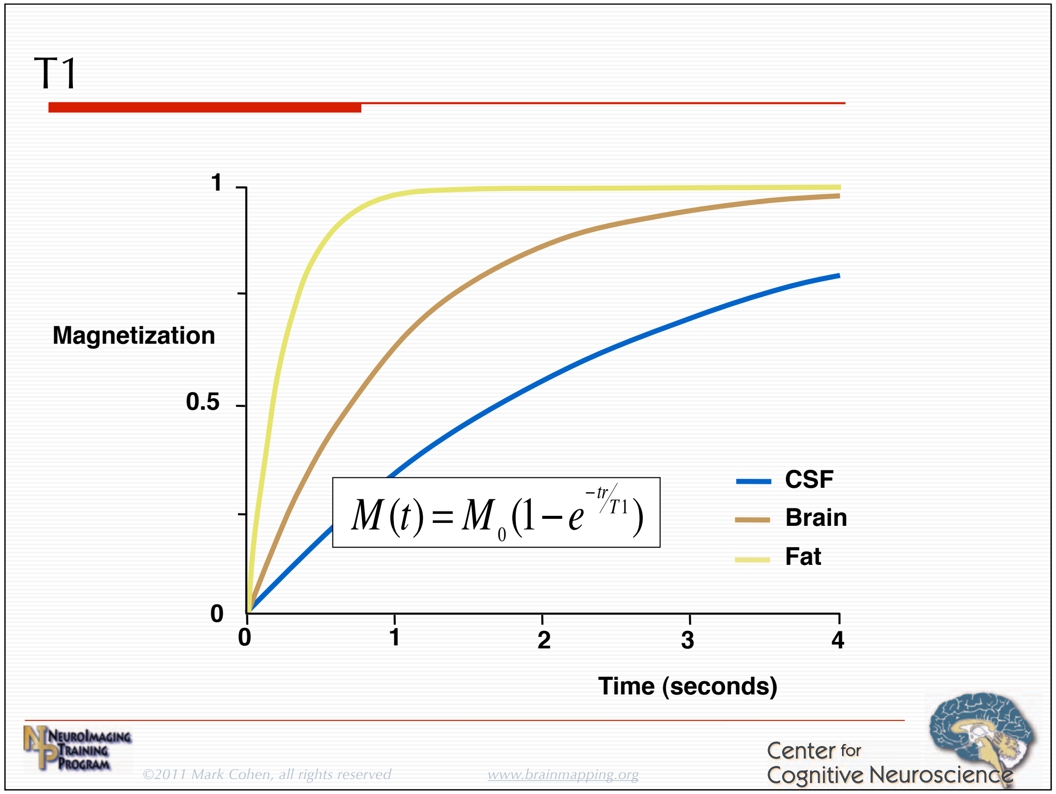
- A Spin 1/2 particle has two states: up and down
- In a magnetic field, B_0, the two states have different energies



When the nuclear magnet is placed into a magnetic field it can adopt only one of two states - aligned with or aligned against - the external field. Then the individual nuclear spins are aligned AGAINST the external field they are in a lower energy state. Moving a system to its lowest energy state is called "relaxation" and amounts to a loss of stored heat energy.



Absent an external field, there will be an equal number of spins in the two allowed states. When an external field is applied, some of the spins change state to oppose it. The local effects on individual spins is much larger than the effect of an external field; therefore only a small number of spins change state (1 in a million or so). The state changes actually require the capture of energy from the environment, which occurs rarely. Therefore, the relaxation to magnetic equilibrium can take some time.



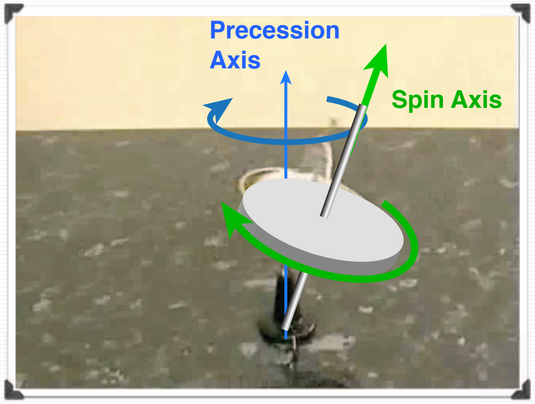
The approach to magnetic equilibrium is exponential. When the system is far from equilibrium, most interactions with the environment will move individual spins to the lower energy state. As the system reaches equilibrium, however, the state transitions become nearly equally likely to be either up or down.

The exponential process is governed by a single time constant, T1, which is characteristic of the tissue type.

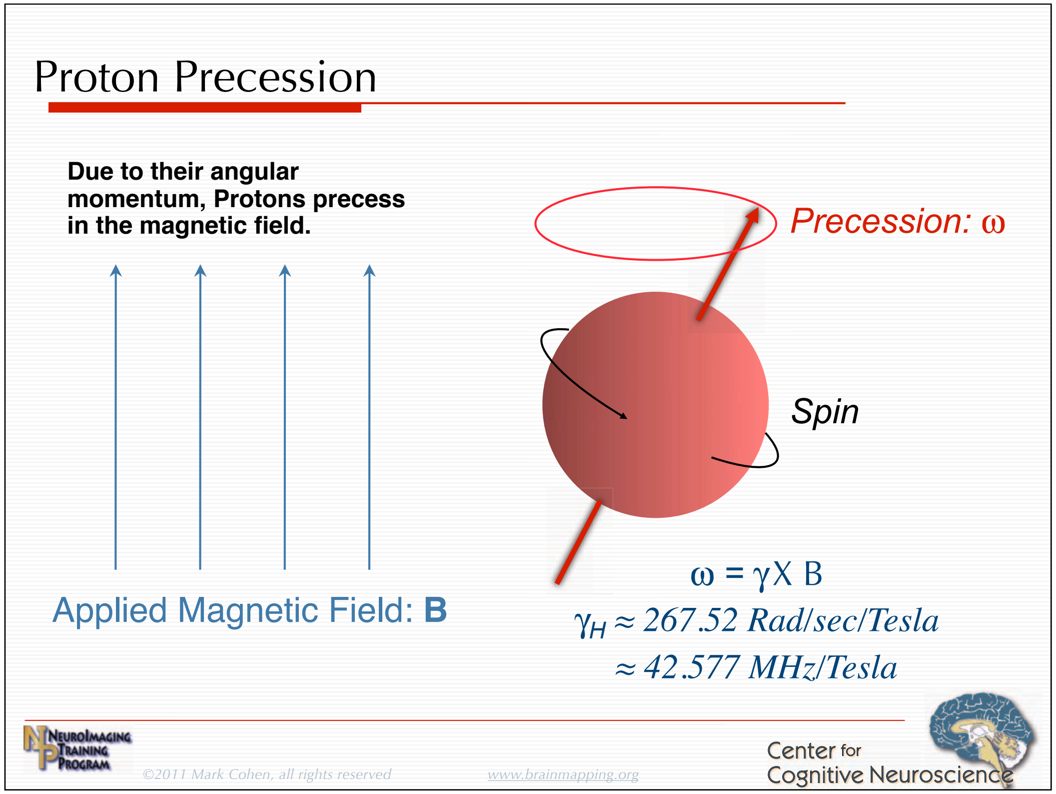
The T1 of water (CSF) is up to several seconds, while the T1 of lipid (fat) is a few tenths of seconds. Most body tissues lie between. The difference in T1 forms a major contrast mechanism in MRI.

Nuclear Spin

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



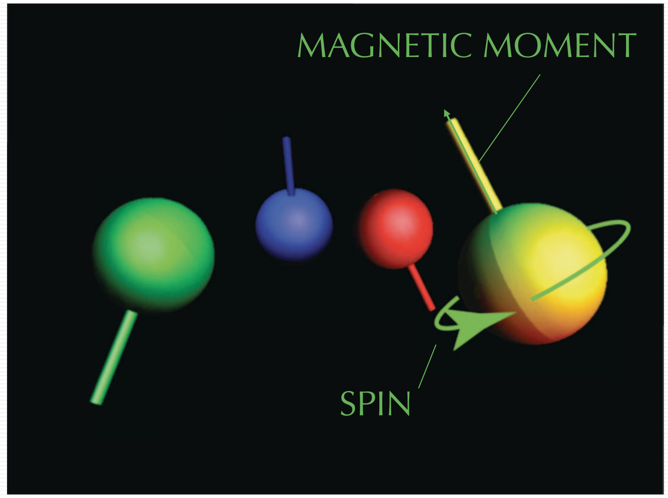
Nuclear spin is a quantized property. The quantum angular momentum behaves like the classical spin of a moving top. Unpaired nuclear spin results in the creation of a magnetic moment, oriented along the axis of spin.



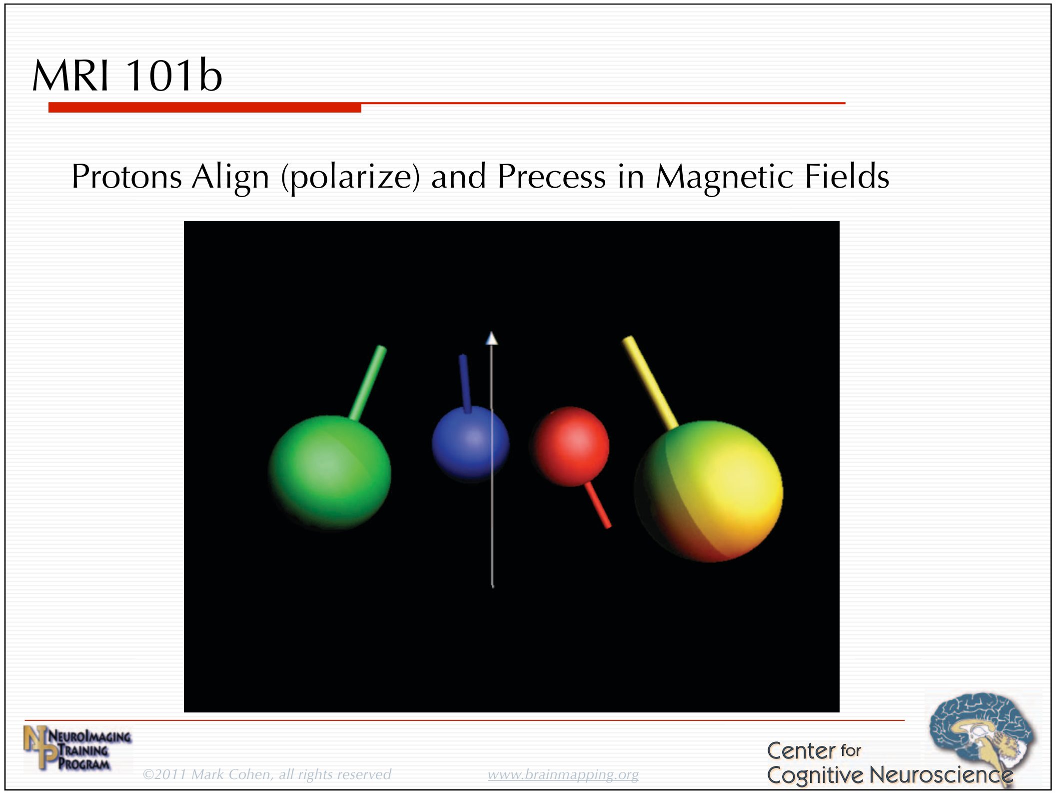
The angular moment of the spins is conserved. This means that when an external field is applied, the individual spins will precess about it. The rate of precession is known as the Larmor frequency and is proportional to the strength of the external field. The "Larmor constant" or "gyromagnetic ratio" is about 42.577 million cycles/second per Tesla.

MRI 101a

Protons, the Nucleus of Hydrogen, Have a Magnetic Moment



To understand how the ultra-low field imager can work, it is useful to review a very few basics of MRI. MRI works by detecting the small magnetic moments of atomic nuclei, principally protons, which are in high abundance in the human body.



When placed in a magnetic field, the protons align with the applied field and “precess” about it. The strength of the alignment, which is called the polarization, determines the strength of the MRI signal that will be received. This polarization shows up as a small excess of protons aligned along the field, rather than aligned against it.

The precession of the protons creates a time-varying magnetic field, that is detected by the imager.

When an ensemble of protons is placed into a magnetic field each of the individual spins adopts the spin up or spin down condition. The precession of the spins also implies that each has a rotational phase. In general, the phases of the individual spins can be expected to be random with respect to one another.

Each spin has a Longitudinal vector component about which it precesses and a Transverse component that rotates about the applied field.

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

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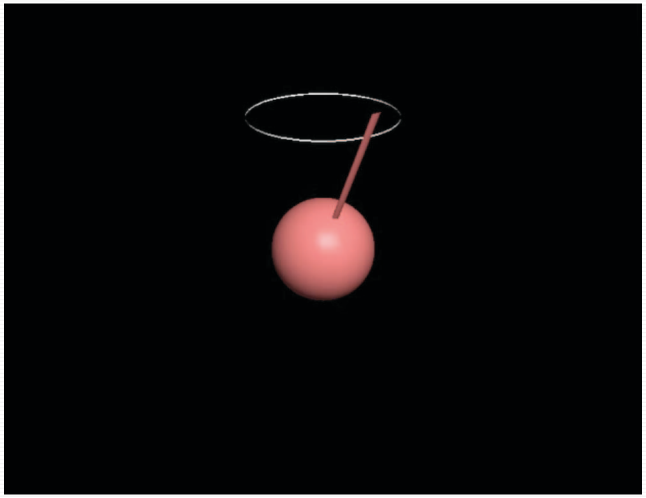
Vector Addition of Spins



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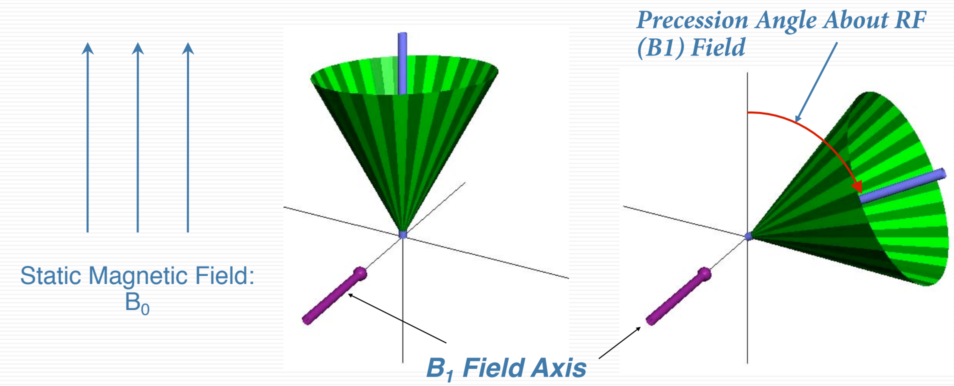
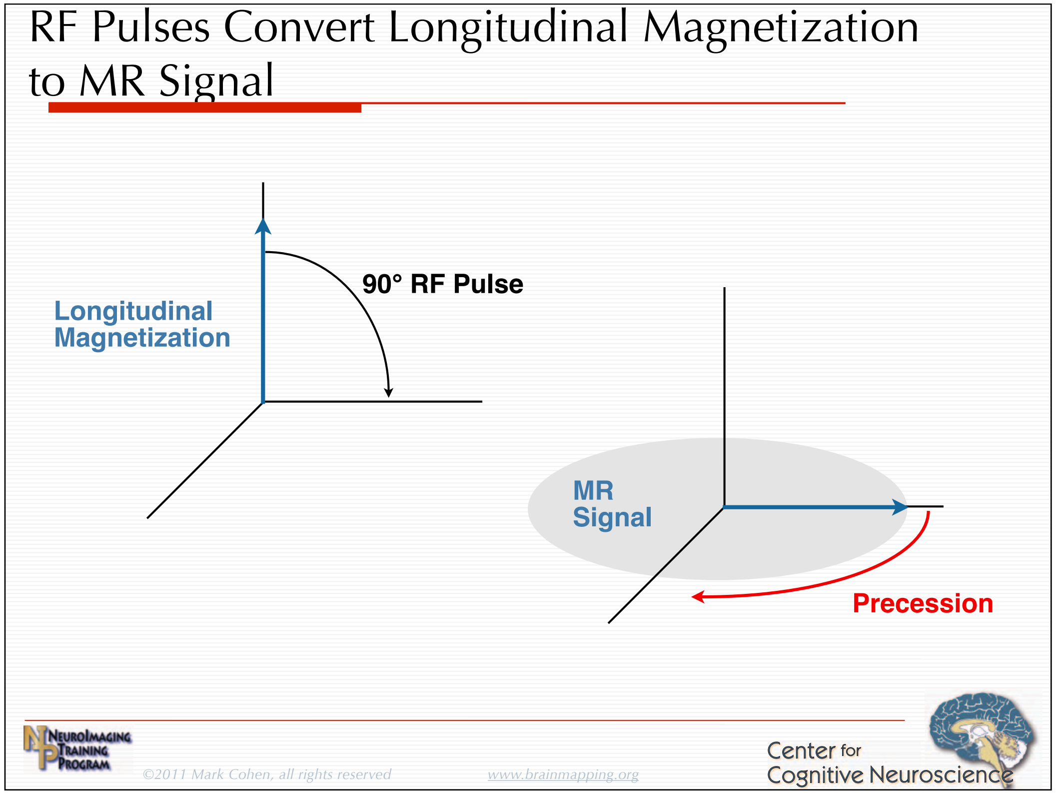
the Resonance Phenomenon



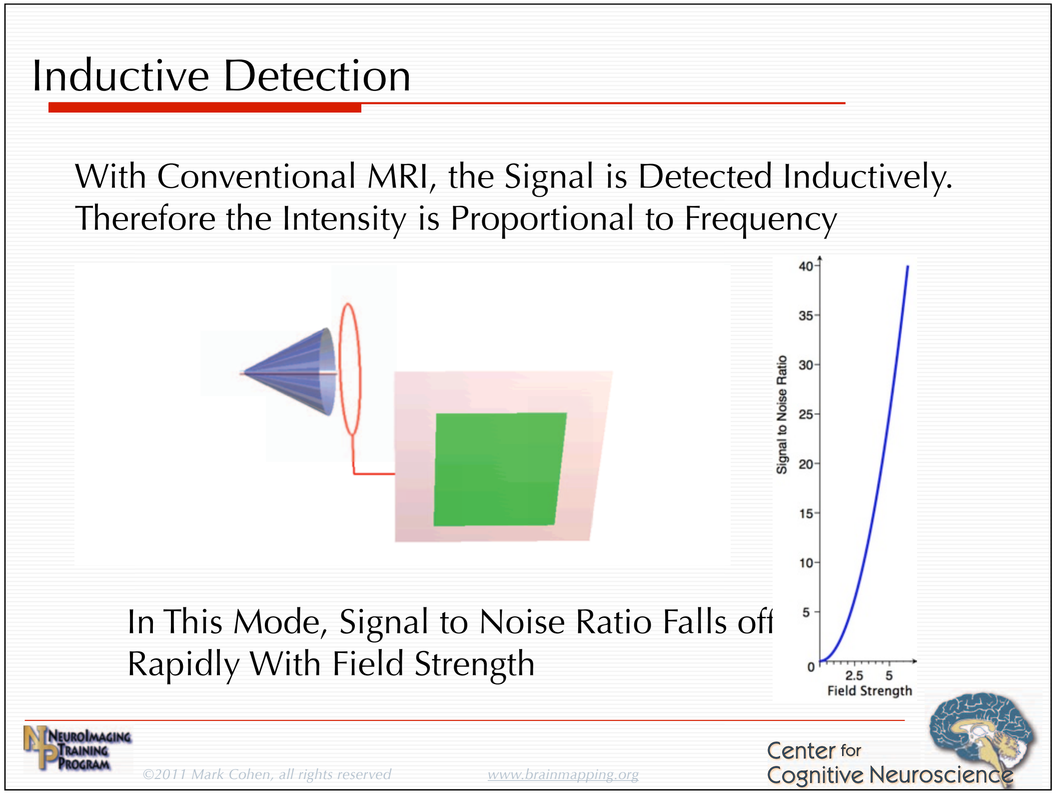
- When a second magnetic field (B_1) is applied, rotating at the Larmor rate, the proton will precess about it.
- The duration and amplitude of B_1 determine rotation angle

If an additional magnetic field is applied the nuclei will precess about their vector sum. In the special case that the second field, B1, is made to rotate at the same (Larmor) rate as the proton nuclei, the B1 field will seem stationary with respect to any invididual proton. They will thus precess about B1 in a simple manner. This view of the interaction of B1 and B0 (the static field) is called the rotating frame. In the rotating frame there is no apparent precession about B0.

Because the Larmor rate at reasonable magnetic fields is typically in the tens to hundreds of MegaHertz, and because B1 is applied just long enough to produce the desired precession away from the longitudinal axis, the B1 field is typically called a Radio Frequency, or RF pulse.



In simple terms, then, the effect of a 90° RF or B1 pulse is to convert any longitudinal magnetization into detectable signal.



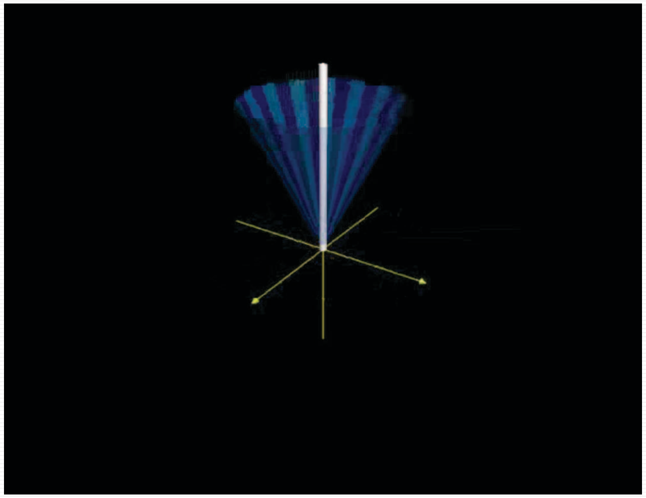
The rotating magnetic field from an ensembel of protons is usually detected by “induction”. The time varying magnetic field from the spinning protons results, by Lenz’s law,

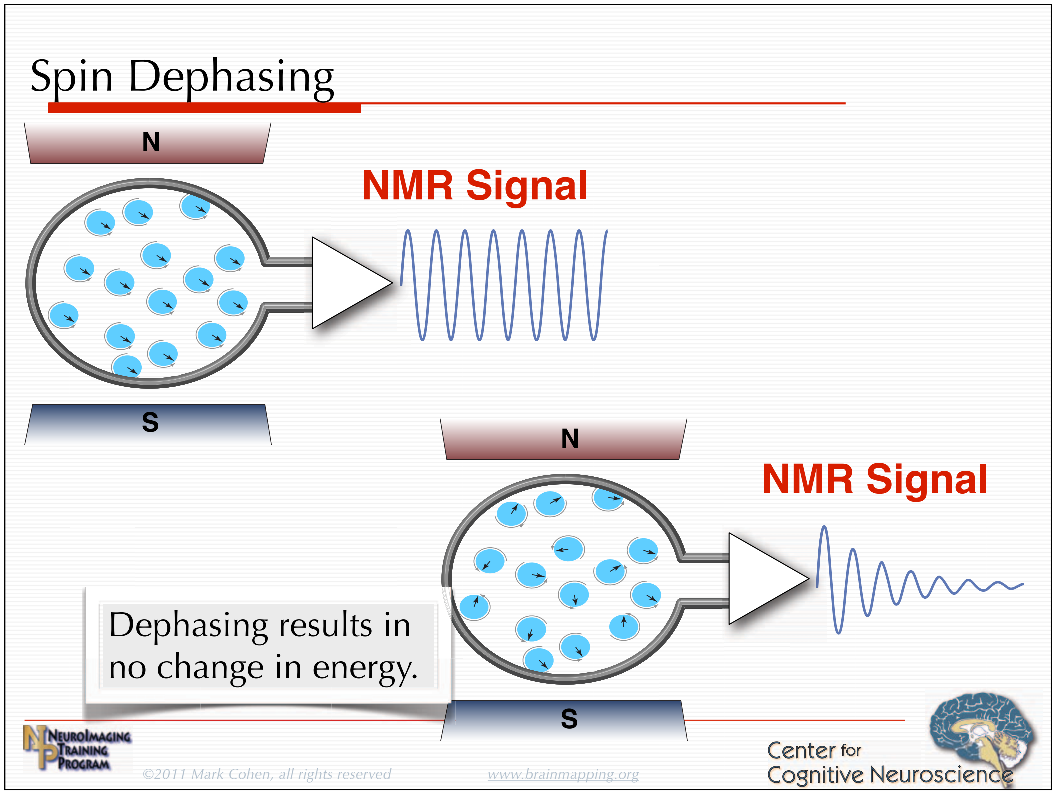
in the creation of a time-varying electic field. This will result in the flow of electrical current in any conductor, such as an antenna. The strength of the electrical field is proprtional to the

Rate of change of the magnetic field. Thus, the detection sensitivity goes up as the magnetic field is increased. For this reason, the signal to noise ratio is a quadratic function of

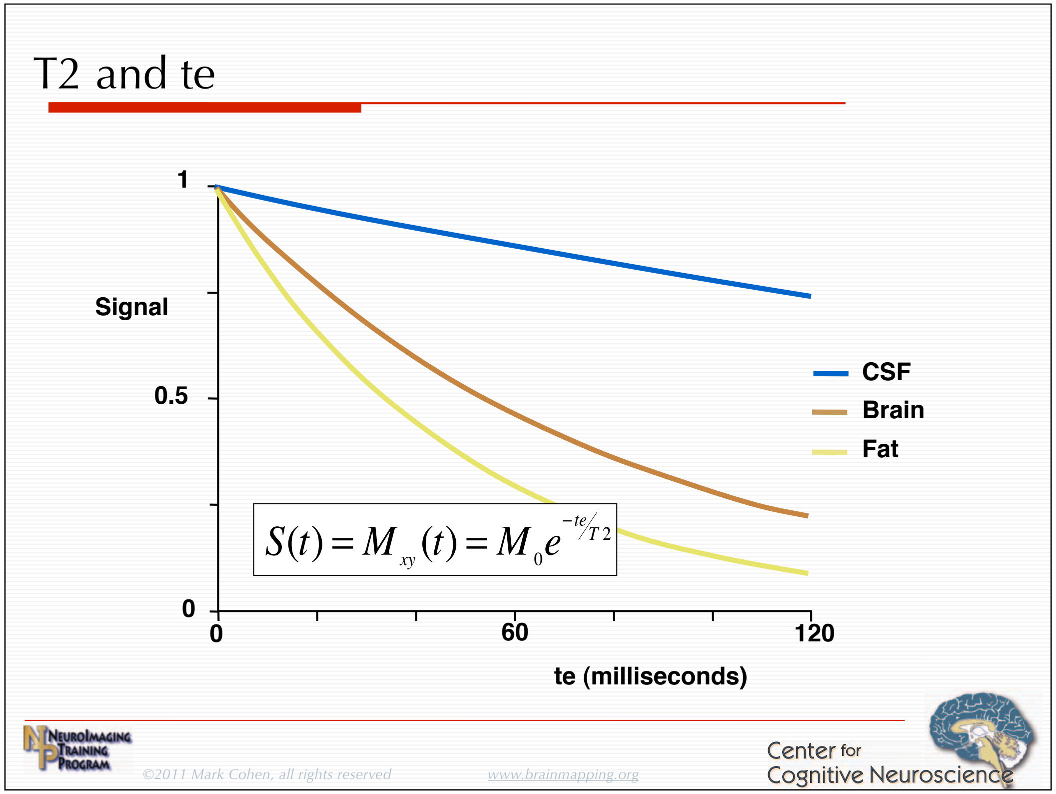
Magnetic field strength, making inductive pickup of MRI signals at bery low field effectively impossible.

From Magnetization to Signal



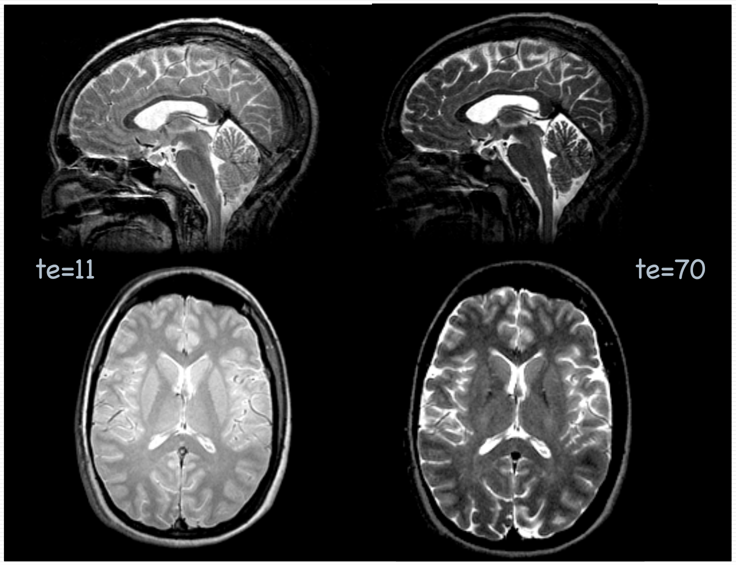


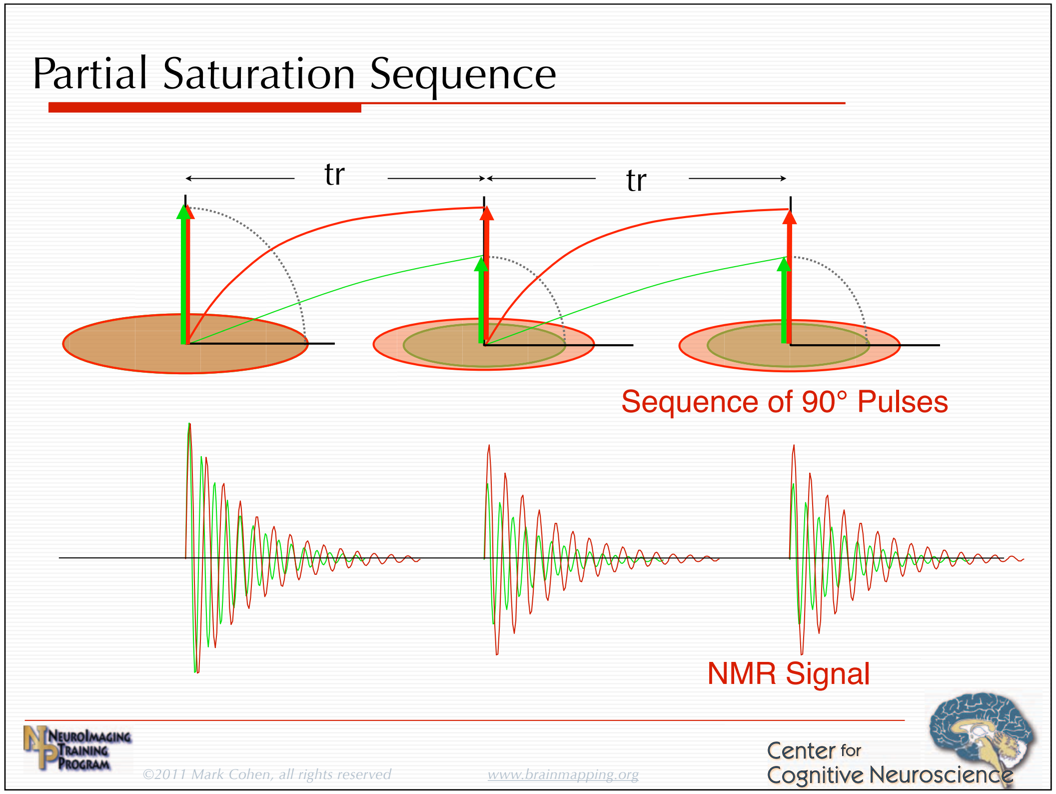
After an RF pulse, the phases of the individual spins will instantaneously be the same - that is, the spins will be precessing “in-phase” and their transverse components will add. This gives rise to a rotating magnetic field that can be detected easily because a time-varying magnetic field creates an equivalent electrical field. If a conductor (antenna) is placed in the vicinity, the electrical field will create a current that can be amplified. Generally, the signal will be a sinusoid at the Larmor frequency.



The MR signal decays exponentially as a result of spin dephasing. The decay is governed by a time constant, T2, that is tissue specific. In the body, the T2 ranges from a few hundred microseconds in bone to hundreds of milliseconds in water (CSF, urine). The adjustable parameter, te, specifies the time after signal excitation at which the data are collected. Longer te’s result in less signal overall, but also in increased contrast between tissues with differing T2

Effects of TE at long TR

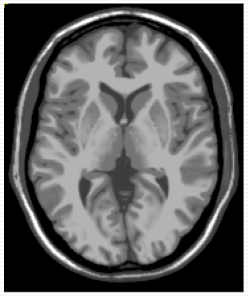




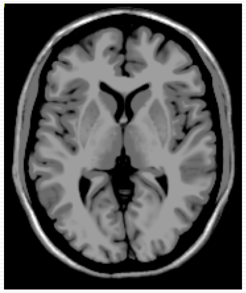
Repeating the excitation and data collection at a fixed interval, tr, introduces contrast based on T1 differences. This is because if the tr is short compared to T1, the spins do not regain their equilibrium state. Each time that a 90° pulse is applied, all of the longitudinal magnetization is lost, and starts again from zero. If a tissue has a short T1, more magnetization is regained, and more signal is created with each 90° pulse. The signal from longer T1 tissues is weaker, however.

tr and contrast (simulations)

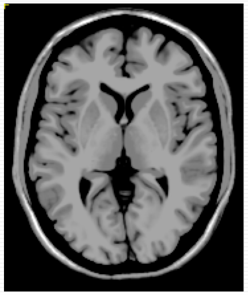
te=4



tr=100

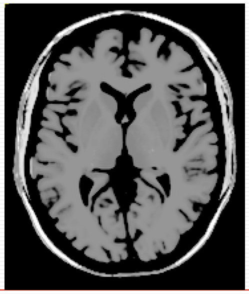


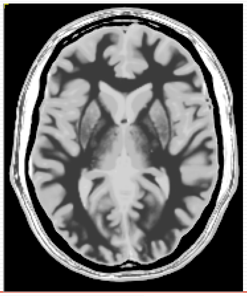
tr=400



tr=700

tr=1500

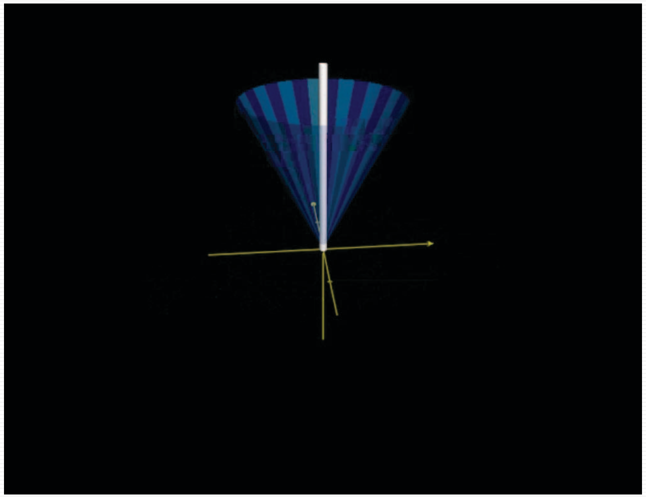


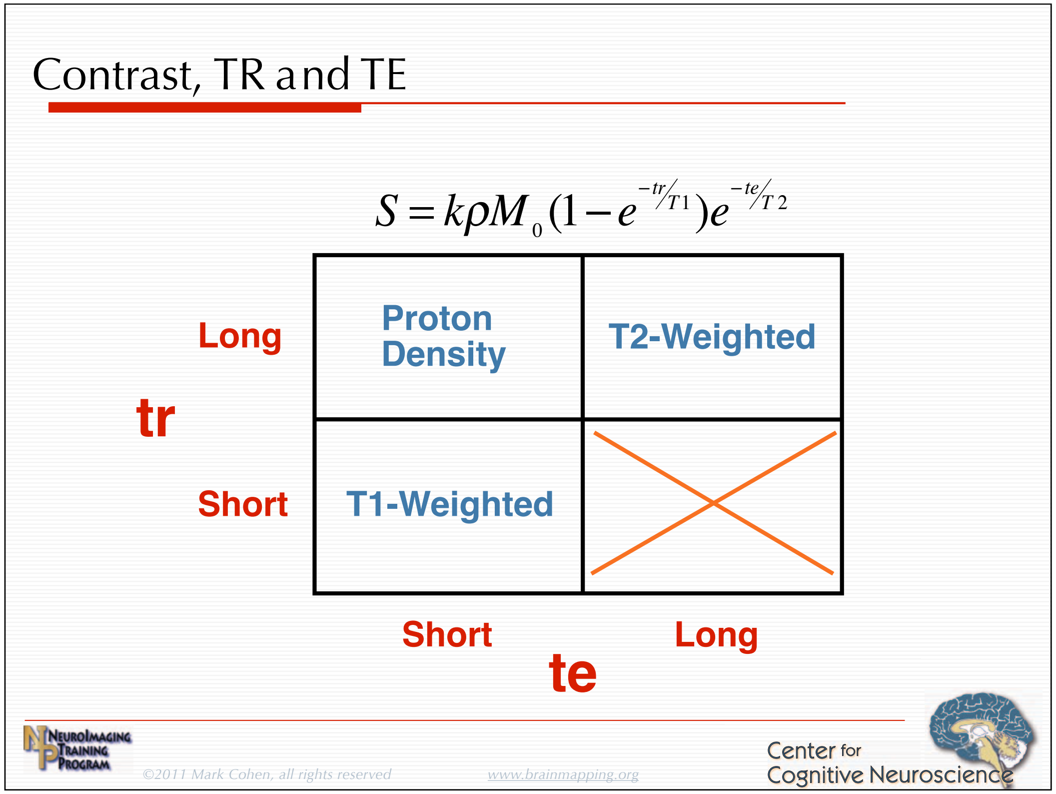


tr=5000

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

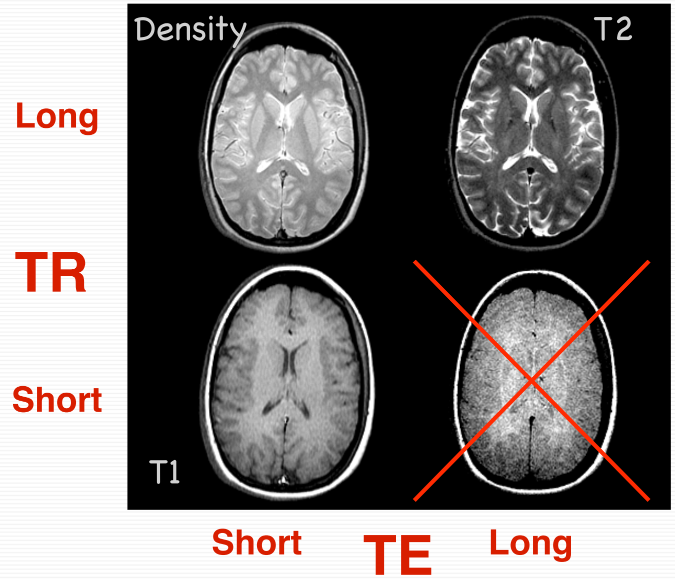
Signal Decay from Dephasing





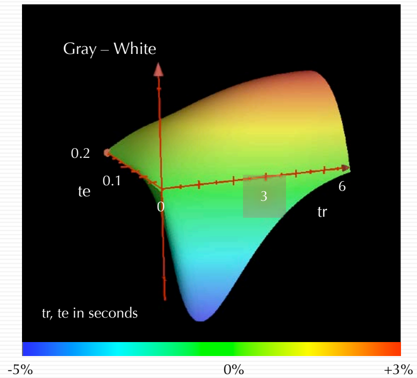
This chart suggests the four principle variations in contrast used in MR imaging. If we consider long tr to be tr >> T1, the images will have little T1 contrast, as all tissues will magnetize almost fully. If tr << T1 the images will have T1 contrast. Likewise, te << T2 minimizes the contrast differences from T2 and te >> T2 maximizes T2 contrast. The equation shows this in analytic form. When tr is long, and te is short, the major contrast determinant would be the density of protons per unit volume. Images with mixed T1 and T2 contrast are typically poor, as shown in the next slide.

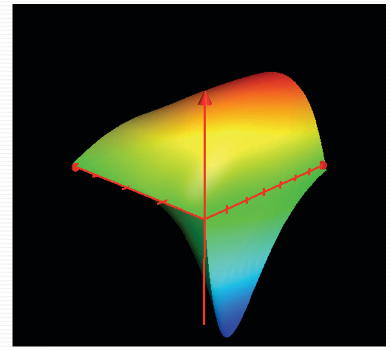
Contrast, TR and TE



Contrast to Noise Ratio

(Gray-White)





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

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Exercise in Afternoon Lab

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The graphs above plot the difference in signal (the “contrast”) between grey and white matter as a function of tr and te. Notice that there is a broad range of combined tr and te where the signals will be isointense (no contrast). The blue regions of the graphs are regions of high T1 contrast, and the red regions have high T2 contrast.

T2*

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

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

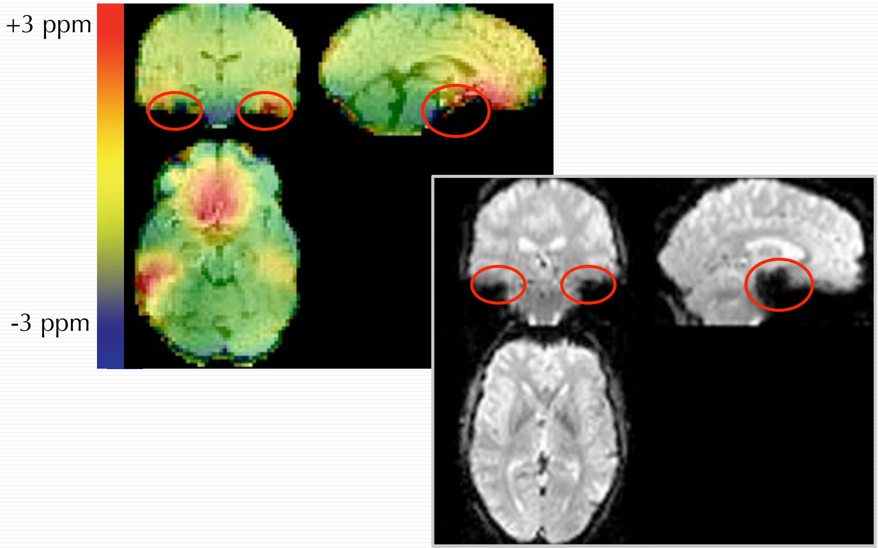
Molecular

Field Inhomogeneity

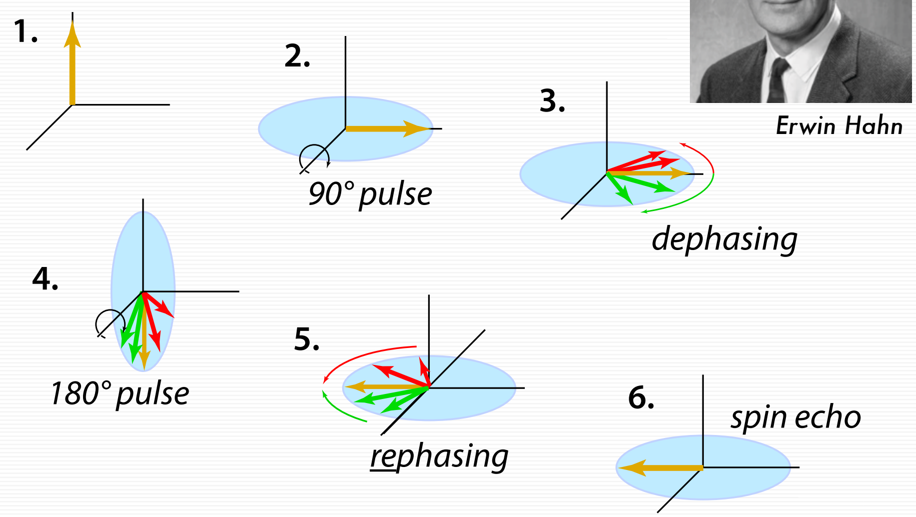
Diffusion

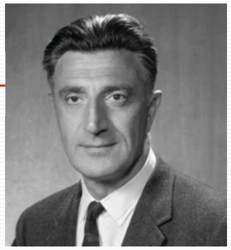
Many physical effects result in dephasing. The process that dephases the spins most quickly dominates the contrast in the final images, thus the net observed T2 (T2*, “T2-star”) is shorter than any of the individual T2’s. Most MR imaging is concerned chiefly with three dephasing effects: molecular interactions (T2), local variations in field from, for example, tissue boundaries (T2’) and dephasing from molecular motion from diffusion (T2D).

Local field Variations Result in Signal Loss

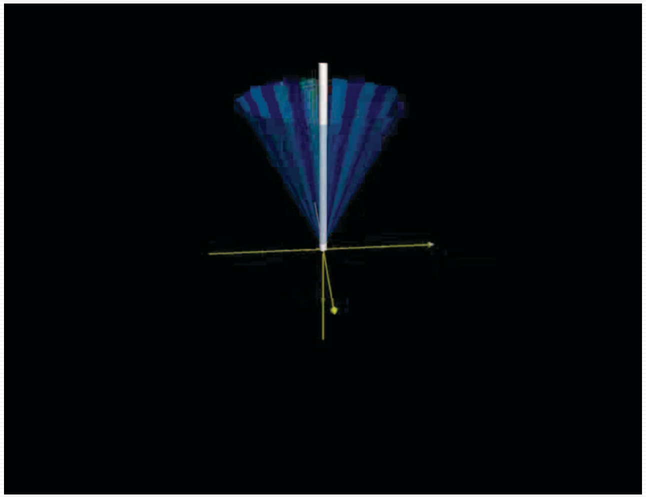


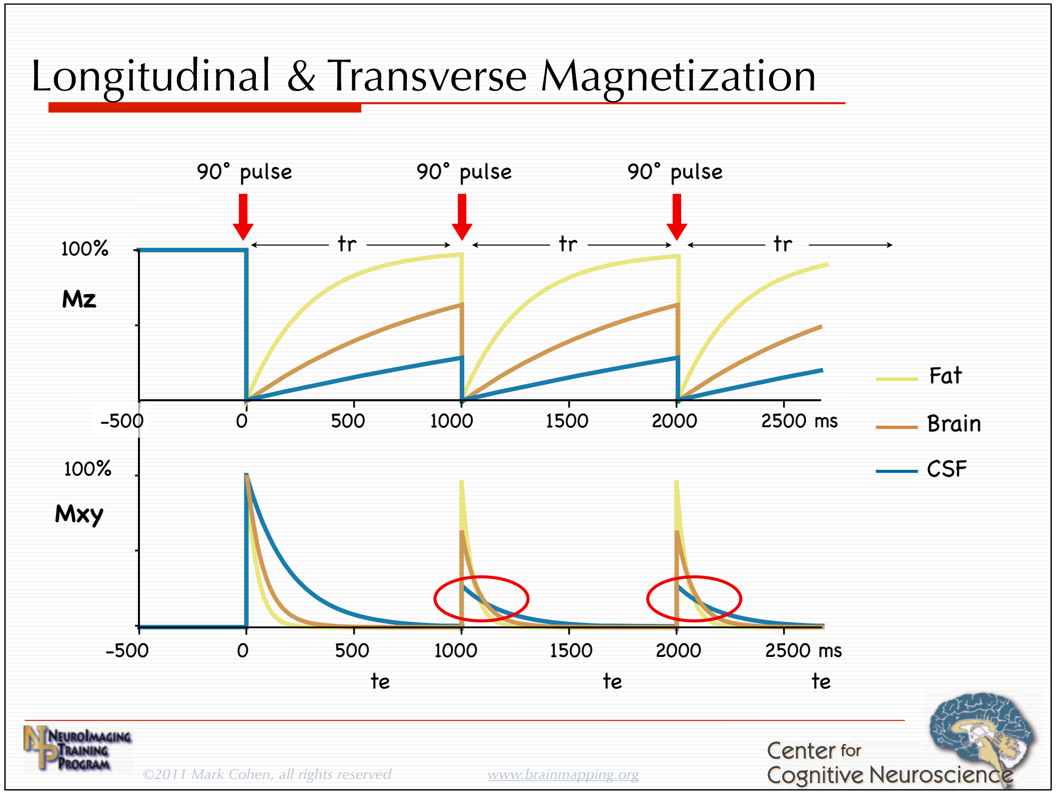
Hahn Spin Echo




Erwin Hahn

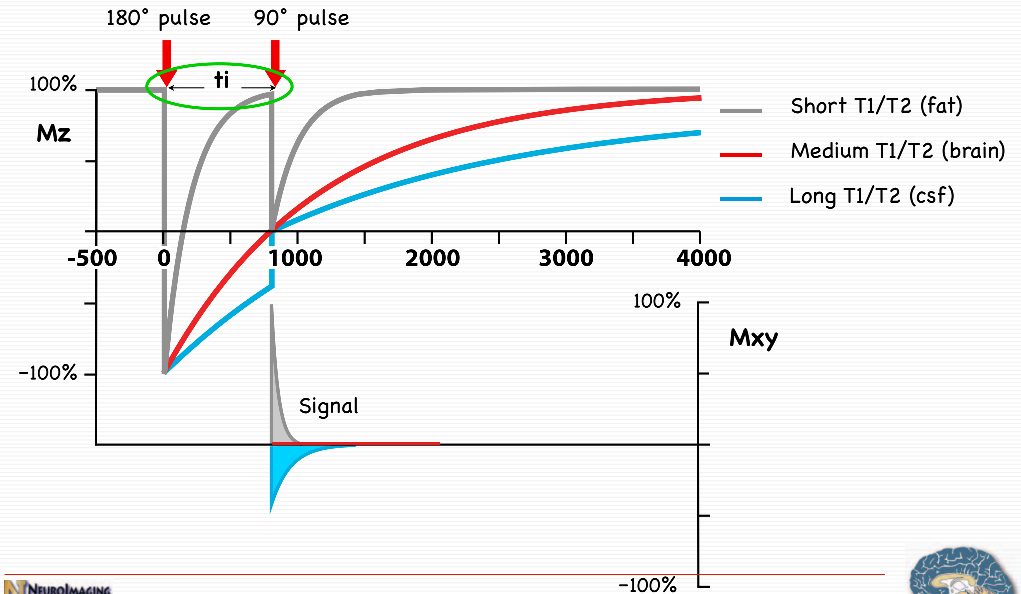
Hahn Spin Echo





This graph shows the combined effects of T1 and T2 on the longitudinal and transverse magnetization during repeated excitation pulses at fixed tr. This model was made with a tr of 1 second, and with typical T1 and T2 of head tissues. Notice that the signals for brain, CSF and fat all cross over with a te of about 70 msec at this tr. The result is that there is no contrast between these tissues. If te is made very short the T1 effects dominate. If te is made very long, the T2 effects dominate, but the signal will be VERY weak.

Inversion Recovery



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

- 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)

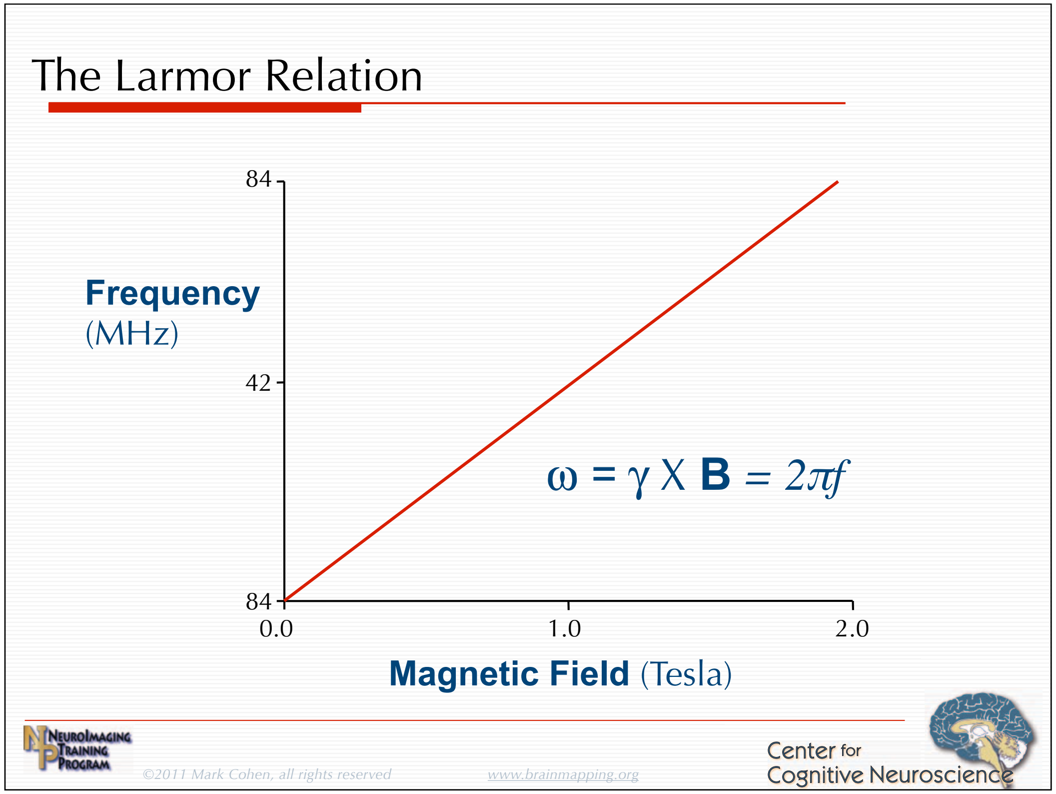
Seventh Inning Stretch

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

- Image Quality and Artifacts (48)

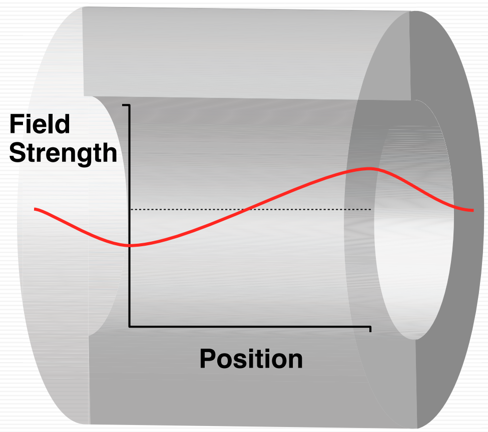
MR Spatial Encoding

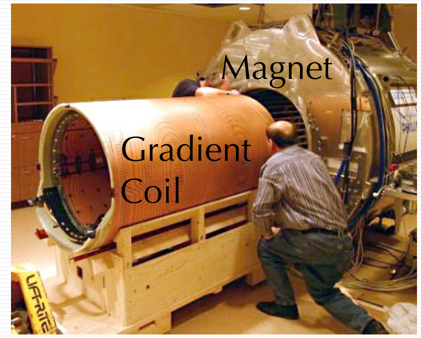
- Most MR Spatial encoding is based on a single concept:
  - If the **Magnetic Field** varies by location, the **MR Frequency** will vary by location,
  - Therefore: We can *find the location by measuring the frequency*.
- Newer methods (e.g., GRAPPA, SENSE) also find location by signal strength as a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



Given what we know about signal intensity and contrast, our next step is to make an image. This implies finding the location of the signal in 3 dimensional space. To do this we rely on the Larmor relation that the precession frequency is proportional to magnetic field. The trick will be to cause the magnetic field to vary with position.

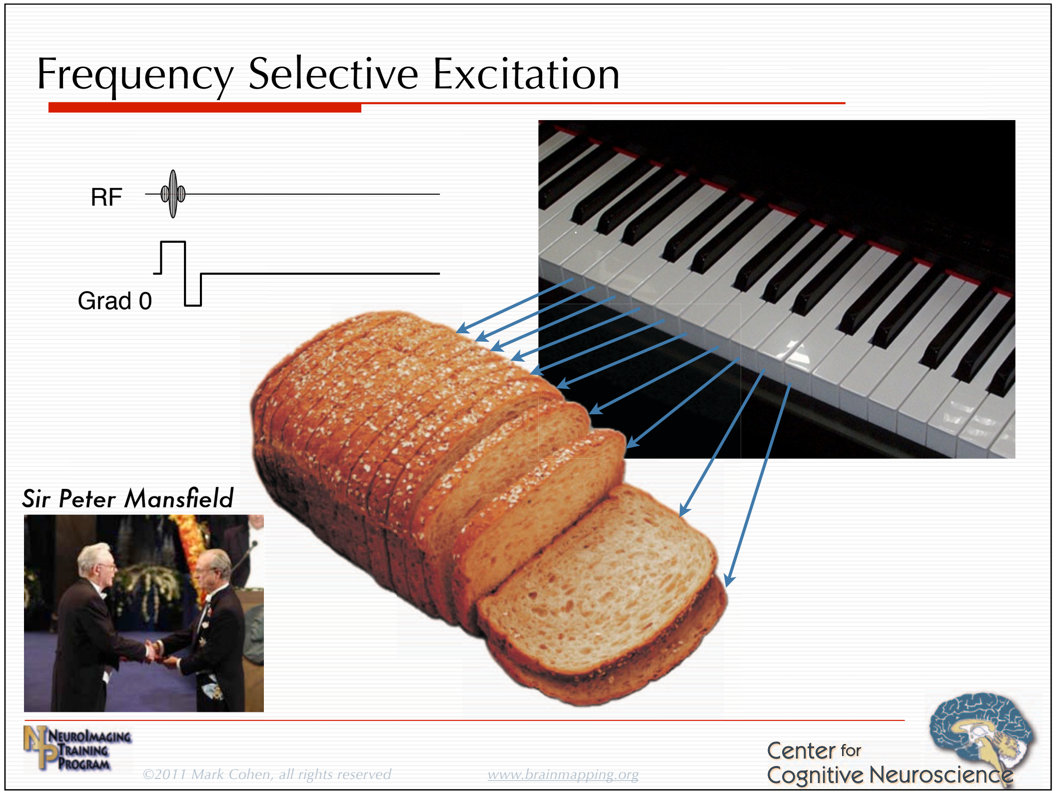
Magnetic Field Gradients





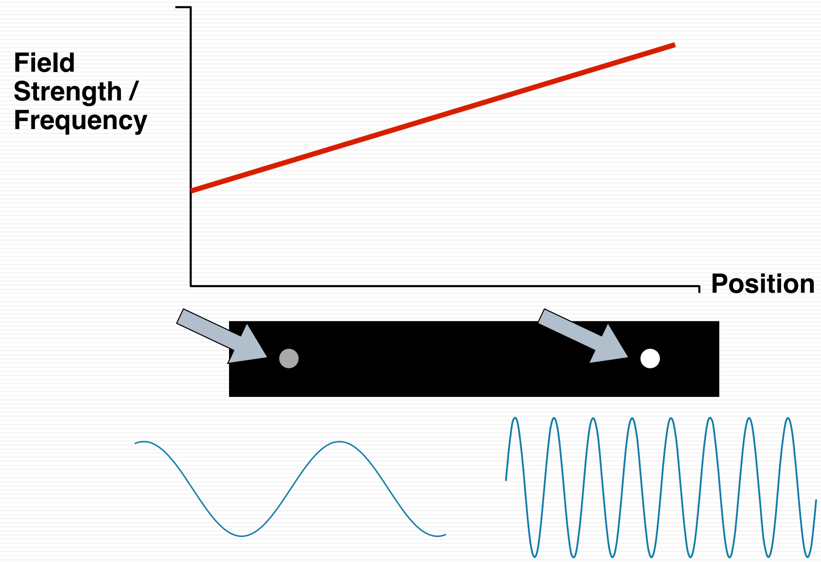
MRI Instrument in Cross Section

We create spatially varying ("gradient") magnetic fields within the bore of the imaging magnet through the use of physical "gradient" coils. The spatial variations are typically small compared to the strong imaging magnetic field. These cause the frequency.



Slice selective excitation is an invention of Nobel-laureate, Sir Peter Mansfield. He noted that if an excitation RF pulse was applied at a fixed frequency in the presence of a gradient field, only spins at the corresponding location in the field would be affected by the pulse and become rotated into the transverse plane (excited). The position corresponds to frequency and can be selected easily.

A 1D Image

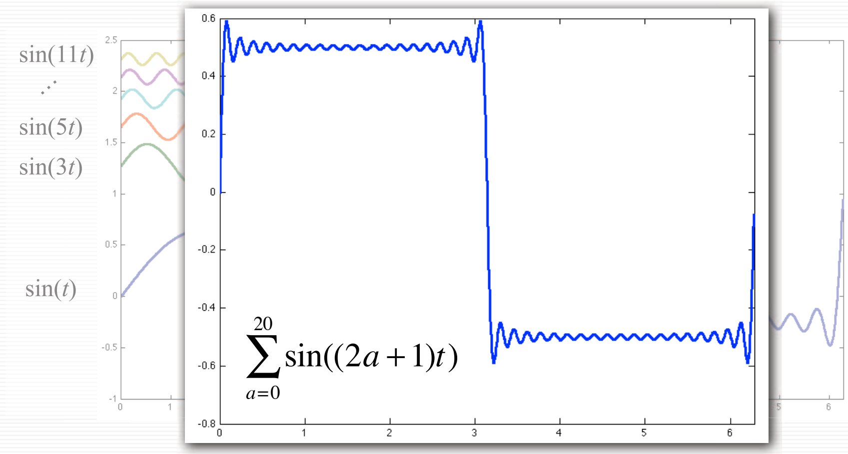


The frequency of spin precession depends on the strength of the local magnetic field. If the field varies with position, so will the frequency. In this diagram, a stronger NMR signal appears in a location having a higher magnetic field.

Interlude: The Fourier Transform

- Any (continuously differentiable) function may be represented as the sum of a set of sinusoids of varied amplitude and frequency.
- The **Fourier Transform** (FT) is “complex”: In general it produces Real (cosine) and Imaginary (sine) components.
- The Fourier transform is computed easily using the Fast Fourier Transform:
  - MATLAB: `>> F = fft(x)`

Fourier Sum example



At left are 6 sinusoids at frequencies of 1, 3, 5, 7, 9 and 11 cycles, weighted by a factor determined by the Fourier transform. At right is the sum of these sinusoids. In the bottom figure, the first 20 odd frequencies are added to make a near perfect ‘square’ function

The Fourier Transform

- The Fourier transform $\mathscr{F}(\omega)$, of a function *f(t)*, is formed from the products of *f(t)* with sinusoids at all frequencies.

The Fourier transform is a function of frequency, ω

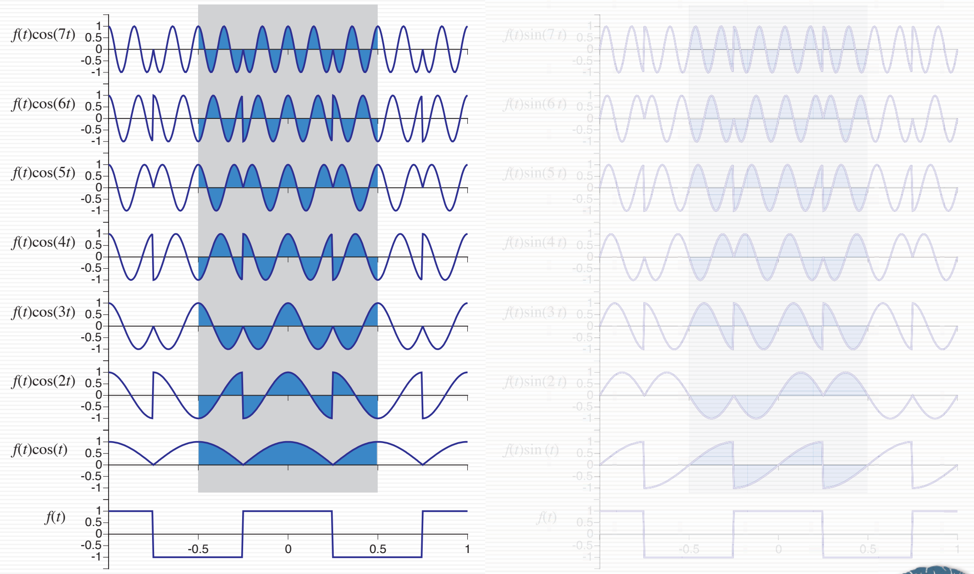
Yikes! $i = \sqrt{-1}$

$$\mathscr{F}(\omega) = \int_{-\infty}^{\infty} f(t)\left[\cos(2\pi\omega t) - i\sin(2\pi\omega t)\right]dt$$

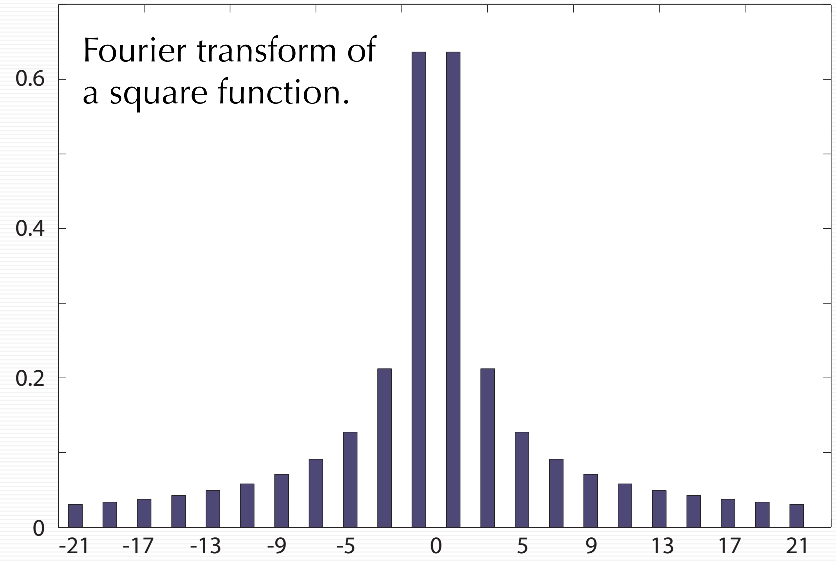
$$= \int_{-\infty}^{\infty} f(t)e^{-2\pi i\omega t}\,dt$$

This amazing result is from Euler
$e^{ix} = \cos(x) + i\sin(x)$

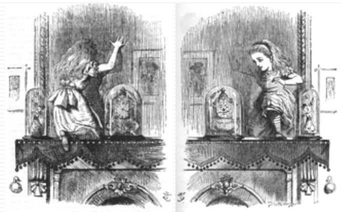
The Fourier Transform is: *the Product of a function and a Series of Sines and Cosines*



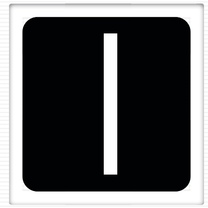
Typical Representation of Fourier Transform

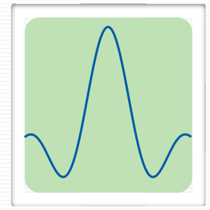


Objects with Finite Width

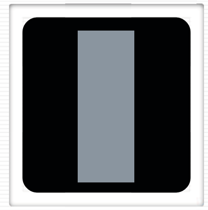


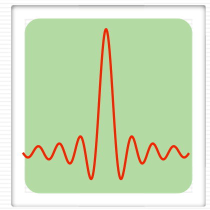
$f(t)$ $F(\omega)$



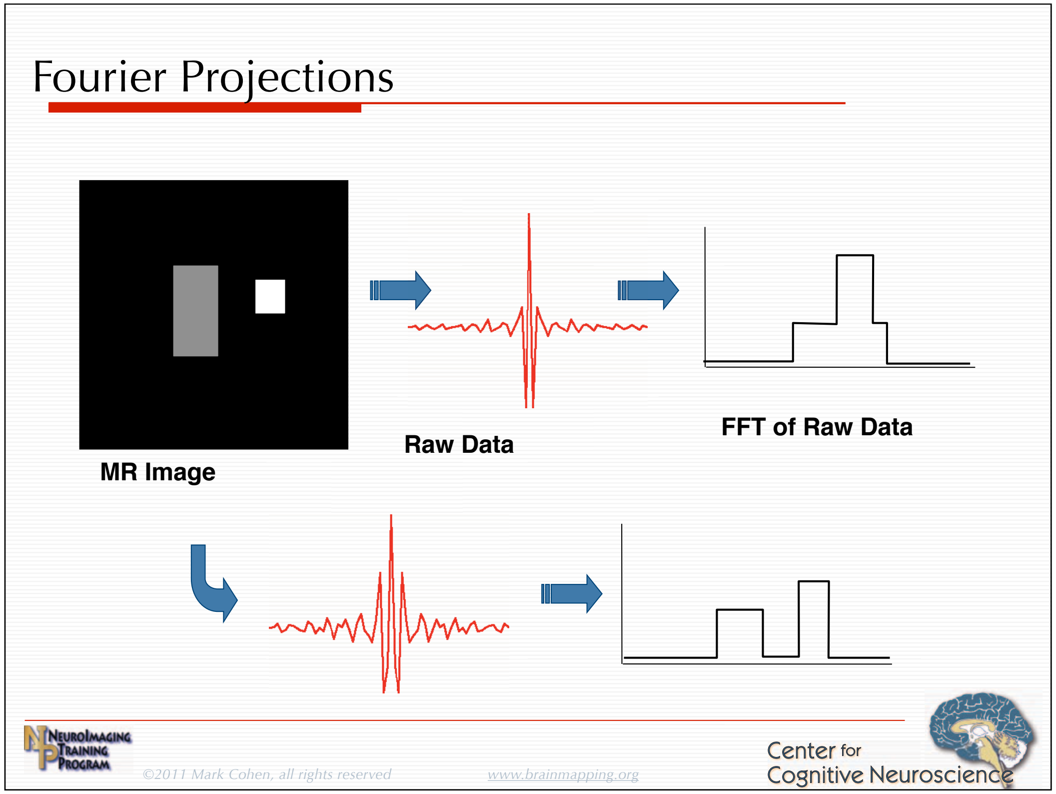


Narrow Feature
➡ Wide Frequency Band

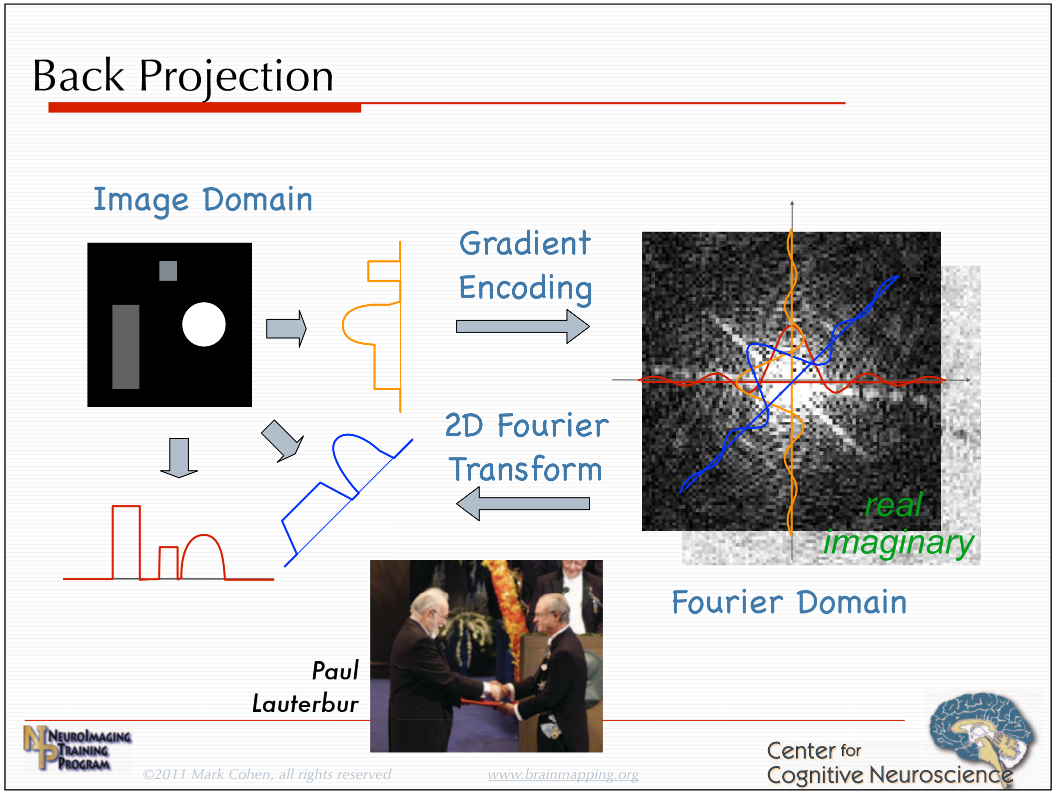




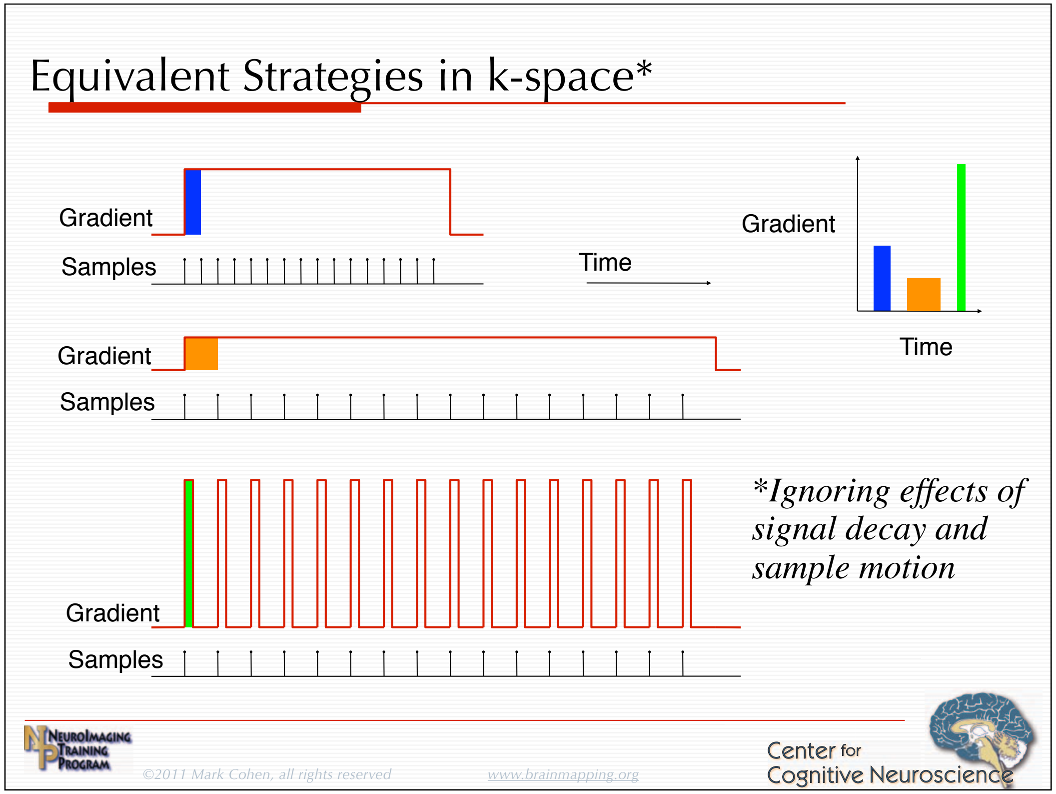
Wide Feature
➡ Narrow Frequency Band



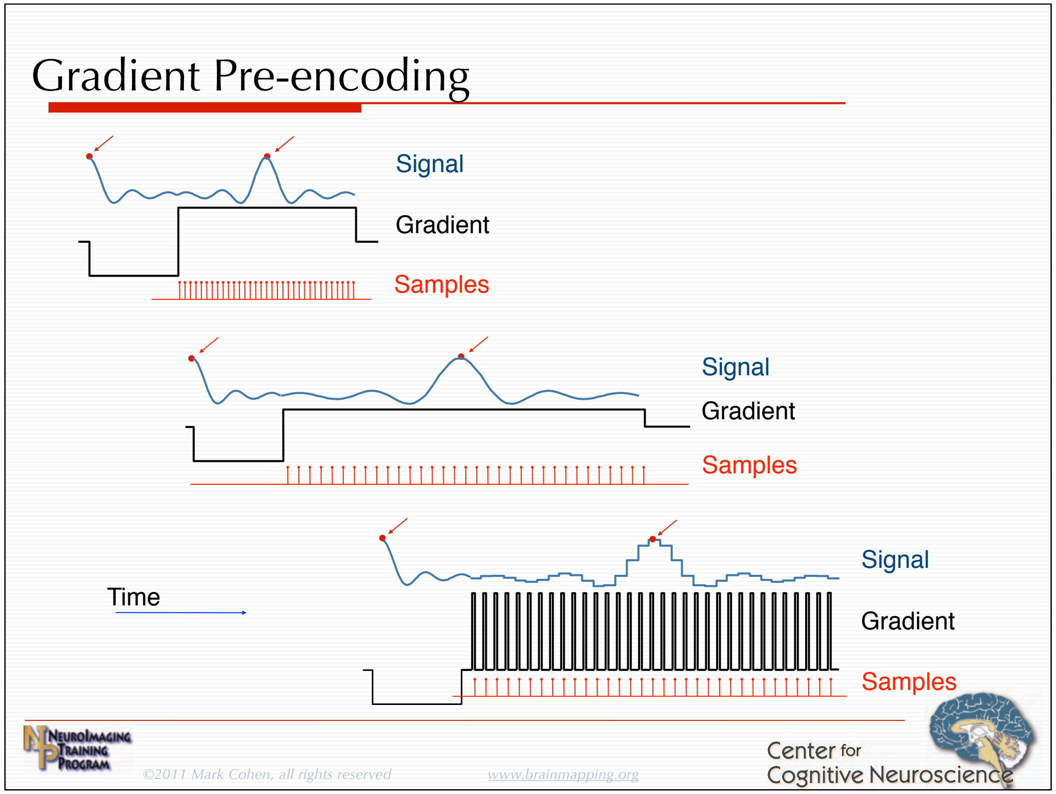
Afer a slice has been excited, we can apply a field gradient in a different direction, causing the spin frequencies to vary along this second axis. The Fourier transform of the signal represents calculates the intensity of the signal at each frequency. As frequency depends on position, we can interpret this as the intensity as a function of location along the same axis as the gradient field was applied. We can readily apply gradient fields along any axis, and thus collect these “projections” in any orientation.



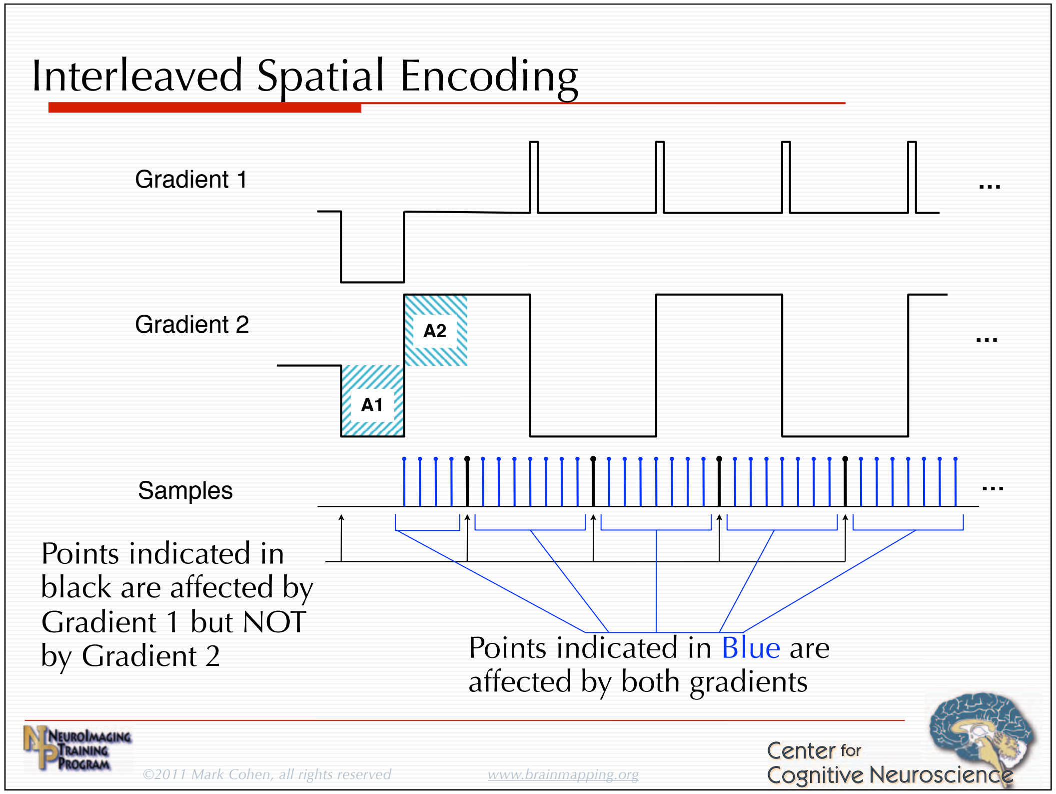
Here we show the projected and transformed signal along three different axes. We can project the ‘untransformed’ data lines into a raw data space according to the direction of the gradient field. The combination of all of these radial projections can looks a bit like rings of waves in a puddle. The 2 dimensional Fourier transform of this combined raw data is the image. A 2D Fourier transform amounts simply to calculating the Fourier transform of the data set line by line, then calculating the Fourier trasform of each column in this intermediate data set.



Crucially, if we ignore all of the contrast effects considered above, the only thing that causes the intensity of the MR signal to change is the effect of the applied magnetic field gradients. While the signal evolves in a continuous manner, in practise, it is sampled discretely and digitized. In the figure, the circles on stems represent sampling time points. In the time between samples (incidentally, this is called the “dwell time”) the differences in precession frequency of spins in different locations causes a phase difference to accumulate as a function of the product of the frequency difference between these locations and the dwell time. Thus, leaving the gradients on at a high amplitude and short duration is equivalent to leaving the gradients on for a longer time at a shorter amplitude. From the perspective of the encoding difference between each sample time point there is no difference. In fact, we don’t have to leave the gradients on for the duration of the dwell interval and instead, we may pulse the gradients at very high amplitude and get the same effect.



The pulsed gradient method (bottom) causes the signal to make discrete jumps in intensity. If the gradients are left on continuously, but the signal is sampled discretely, we collect identical data. Note in these figures that the spins are typically “pre-encoded” by applying the field gradients in their opposite sense, before collected the signal. This ensures that all of the spins will be in-phase at the center of the readout period.

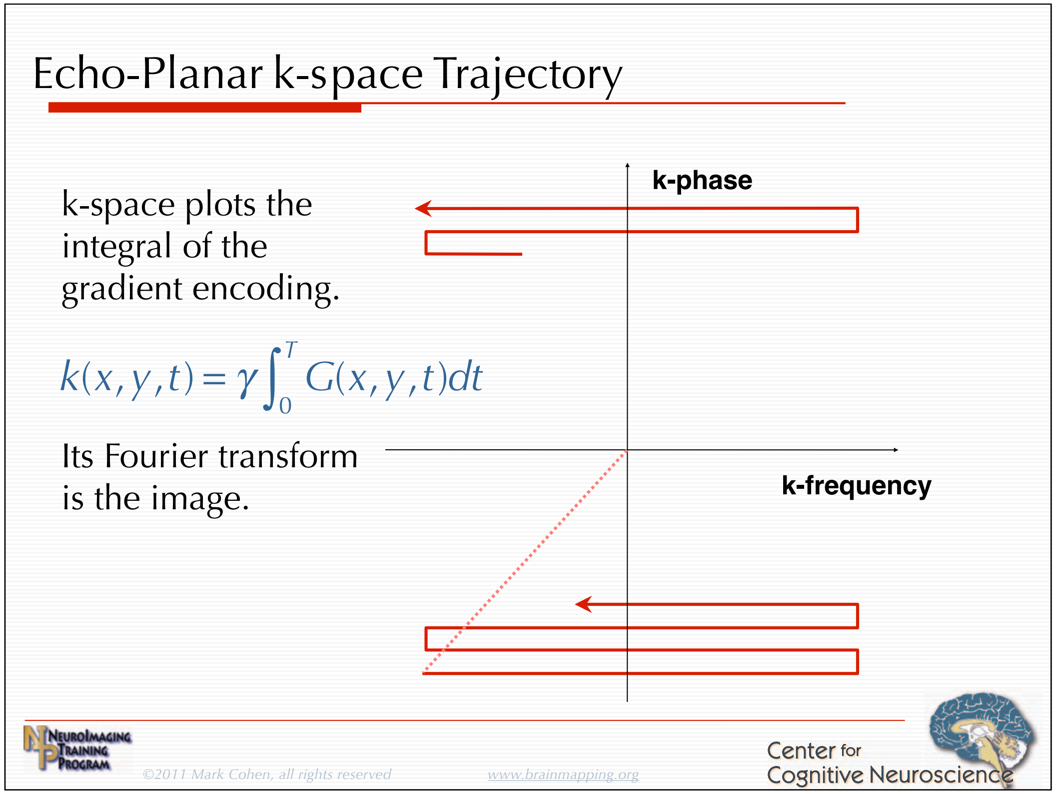


As a gradient is applied for a time period (A1) the spins in different location go out of phase with respect to one another. When the gradient is applied for the same duration in the opposite sense - reversing which side of the instrument is at higher field and which is lower - the spins go back into phase.

Considering only Gradient 2 above, at the time points indicated in black, the spins are always in phase. Alternating the sign of Gradient 2 makes it invisible at these time points. However, if an orthogonal gradient, Gradient 1 is applied between these oscillations, it will cause a phase accumulation along the Gradient 1 direction.

For example, if Gradient 1 is directed along the vertical (Y) axis of the magnet, the time points indicated in balck represent a line with spatial encoding along this axis only. The Fourier transform of this, of course, is a signal intensity projection along the Y axis. The points indicated in blue are affected by both gradients and fill in the other points in the 2D raw data space.

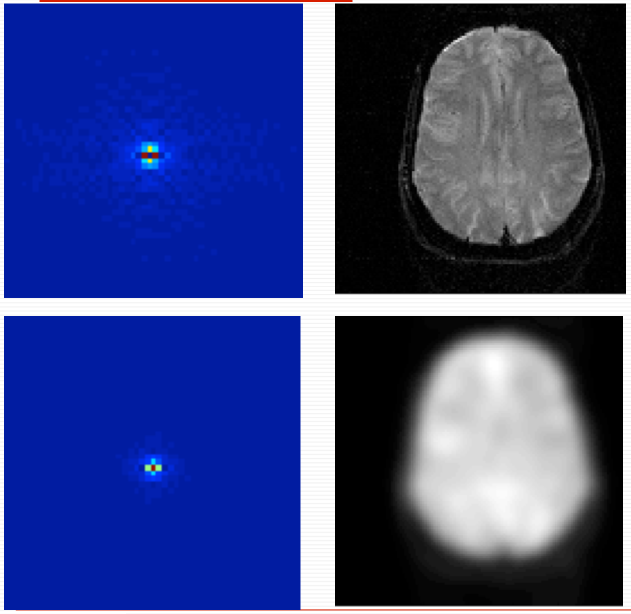
In this way, we can interleave the spatial encoding in the X and Y axes creating a 2D image all at once. This scheme is known as Echo Planar Imaging or EPI and is the fastest practical method to form MR images at this time.



If we examine the locations of the raw data points that are collected in EPI we see that they are collected line-by-line in alternating directions. The brief pulses of Gradient 1 move the locations from one line to the next. The net effect is to make a raster pattern in the raw data space. This raw data space is known as “k-space” and is the 2D Fourier transform of the image. As long as there are no other things changing the signal, the order in which the data are placed in k-space makes no difference.

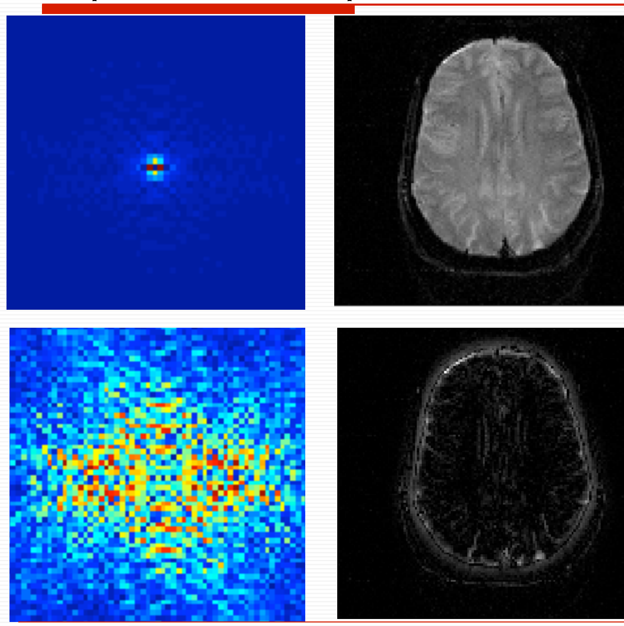
Since the amount of spatial encoding between spins is determined by the product of the gradient amplitude and duration we generalize this as the integral of the gradient-time product. This product becomes the phase difference between spin locations; k-space is sometimes referred to as phase space for this reason.

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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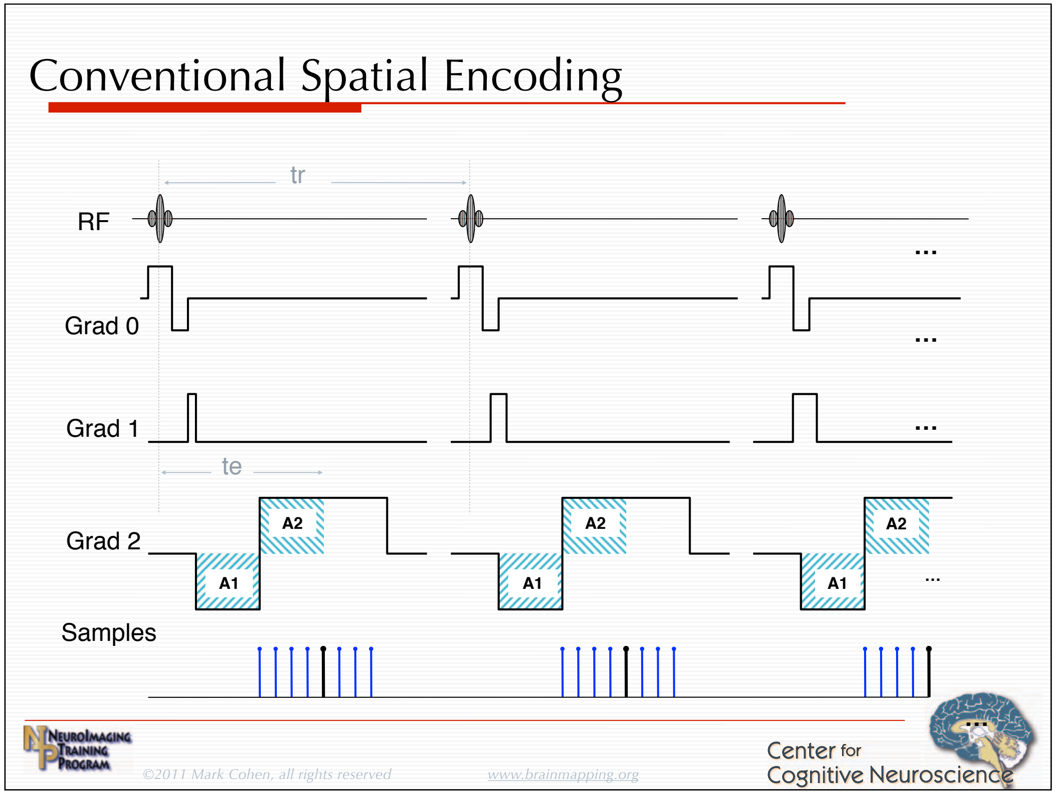
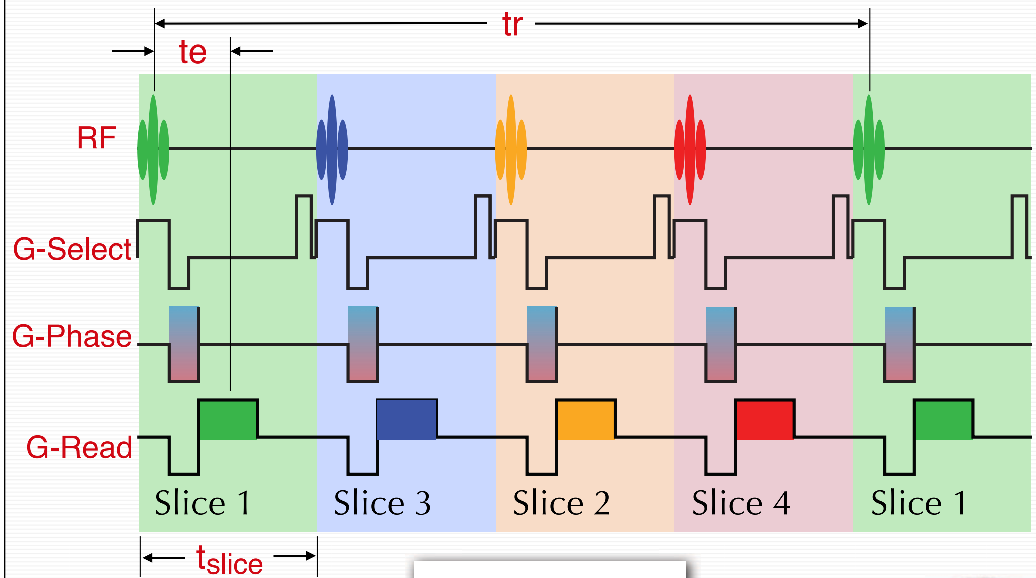
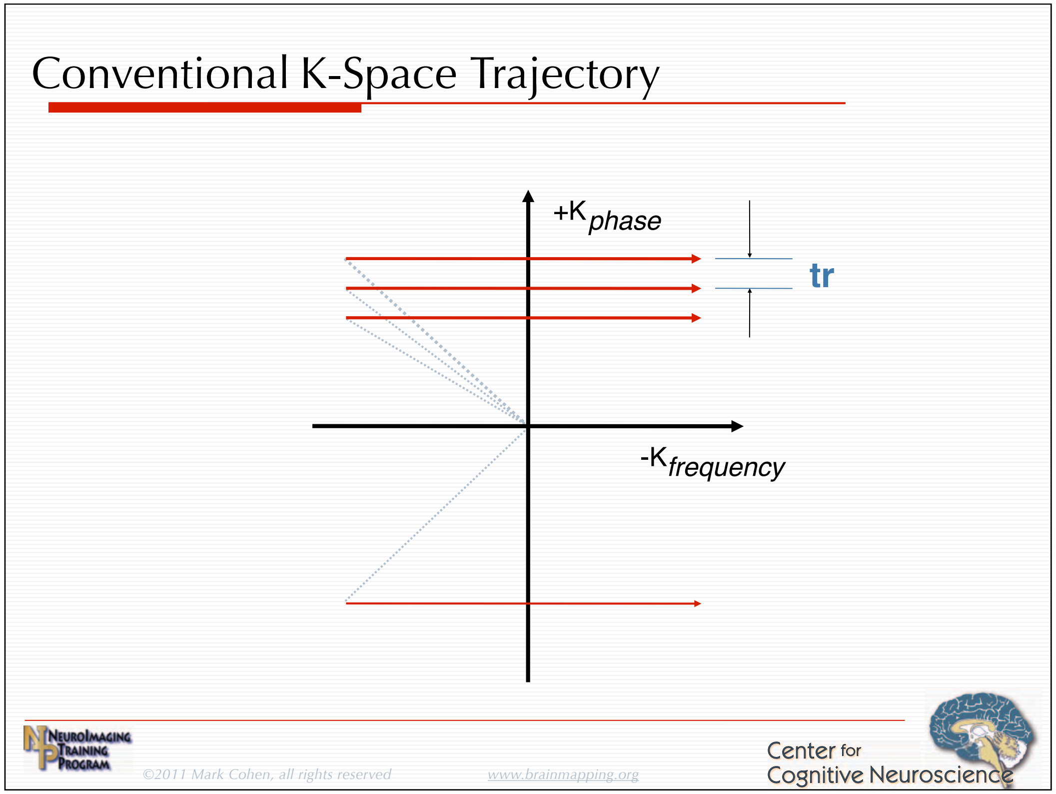


Before echo-planar imaging became possible, a different encoding scheme was used that collected only one line of k-space with each tr. The position along the y-axis of k-space is established by applying a brief pulse of the Y-gradient (Grad 1) before collecting the data in the presence of the X-gradient. With each tr, the area of the Y-gradient pulse is increased. The diagram above also shows the selective excitation pulse, wherein an RF pulse is transmitted in the presence of a gradient directed along the z-axis (Grad 0). Patterns of this kind that indicate the timing of events in MRI are known as pulse sequences.

Multi-slice MRI

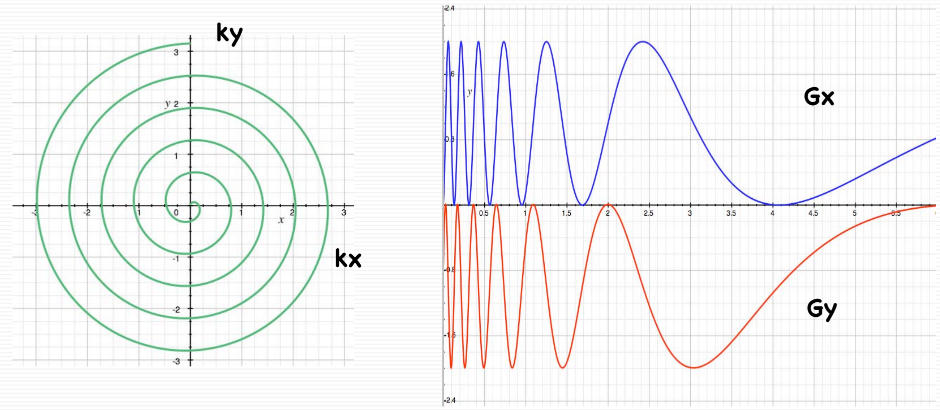


$N_{slices} = tr / t_{slice}$



The traditional line-by-line manner of forming MR images results in the k-space filling pattern shown above. Note that time does not appear explicitly in the k-space diagram. In the conventional imaging case, the total imaging time is determined by the number of lines of resolution in the Y or phase encoding axis and by the time between lines, which is tr.

Spiral



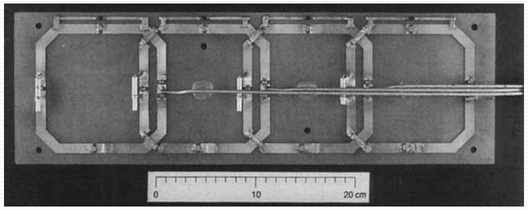
NeuroImaging Training Program

www.brainmapping.org

Center for Cognitive Neuroscience

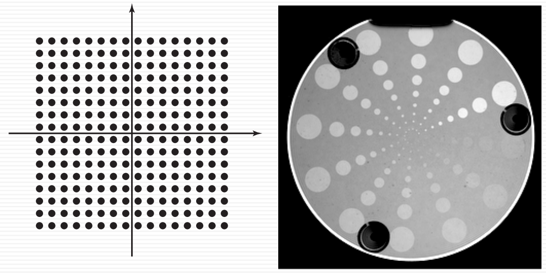
The order in which k-space is populated is somewhat arbitrary. One filling pattern, “Spiral Scanning” has significant popularity. The idea is to make a spiral trajectory in k-space. It is easy to see that this requires a gradient pattern in X and Y that is a slowing sinusoid.

The NMR Phased Array

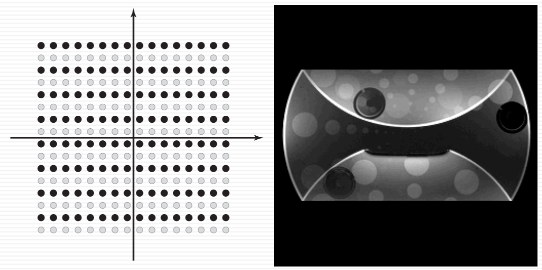


Peter Roemer. et al.,
Magn Reson Med. 16:192, 1990

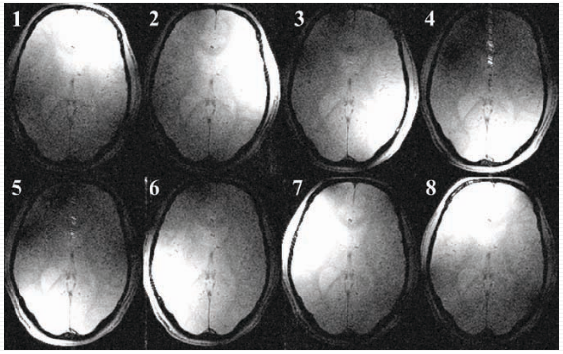
SENSE Encoding



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



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


1
2
3
4
5
6
7
8

Spatial Encoding Summary

- Spatial Encoding and Contrast are Linked through the Pulse Sequence;
- The MRI Raw Data are the *2D Fourier Transform* of the Final Image (usually the *Magnitude Transform*);
- Spatial Encoding is Added through Gradient Coils;
- 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

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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…

NeuroImaging Training Program

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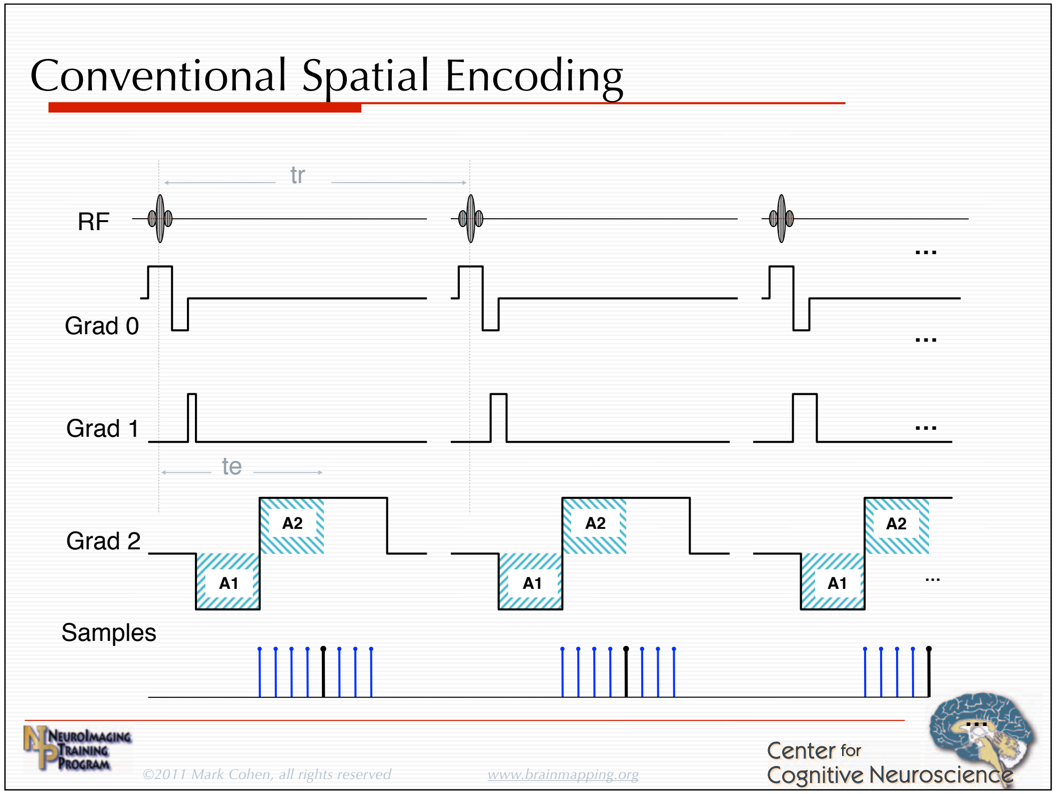
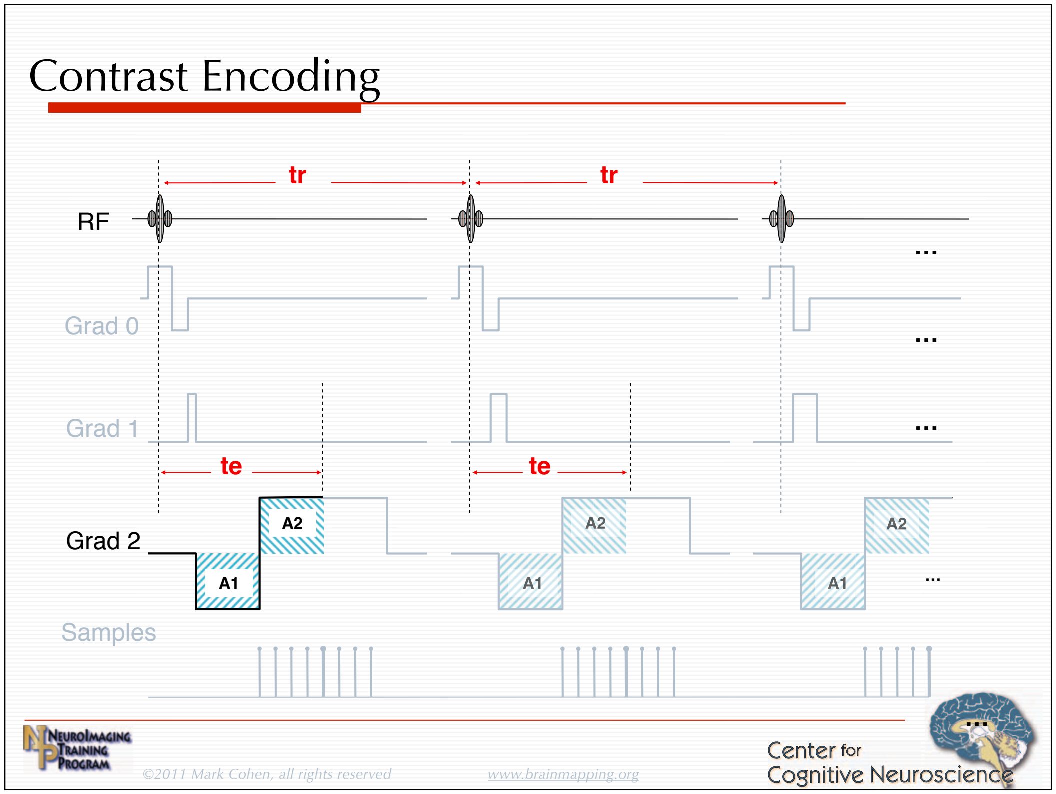
Center for Cognitive Neuroscience

The pulse sequence ultimately controls a host of factors. The user seldom interacts directly with the gradient and RF pulse waveform controls, but instead enters parameters for acquisition such as the slice locations resolution features and contrast controls.



Before echo-planar imaging became possible, a different encoding scheme was used that collected only one line of k-space with each tr. The position along the y-axis of k-space is established by applying a brief pulse of the Y-gradient (Grad 1) before collecting the data in the presence of the X-gradient. With each tr, the area of the Y-gradient pulse is increased. The diagram above also shows the selective excitation pulse, wherein an RF pulse is transmitted in the presence of a gradient directed along the z-axis (Grad 0). Patterns of this kind that indicate the timing of events in MRI are known as pulse sequences.



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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

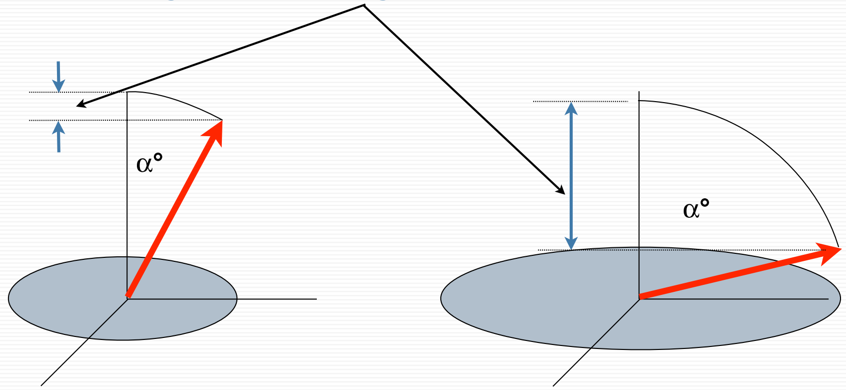
TR and Image Quality

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

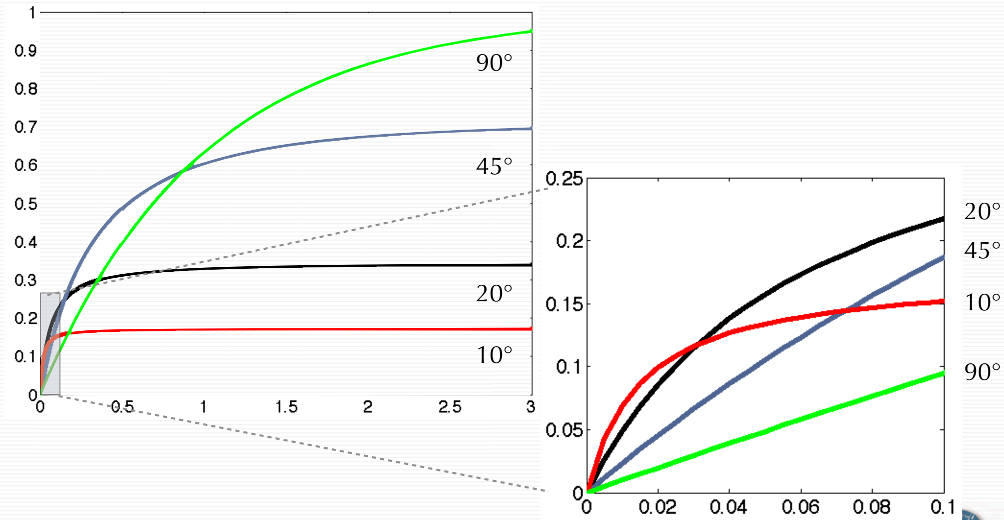
Loss of Longitudinal Magnetization



Small Flip Angle

Large Flip Angle

Flip Angle and TR/T1



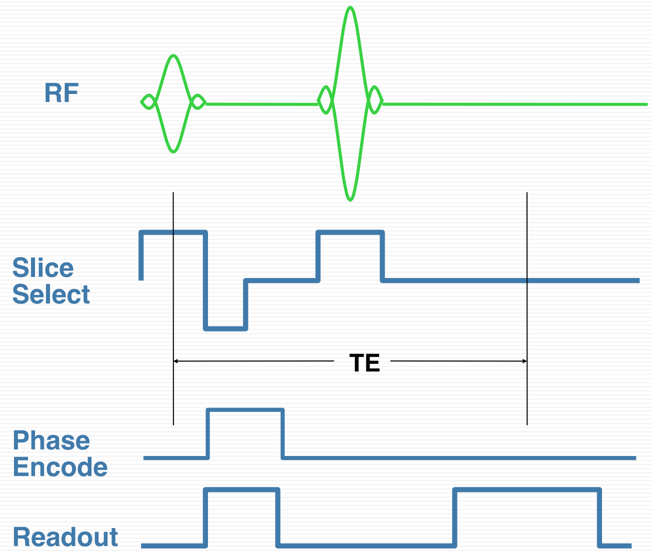
Contrast and Flip Angle

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

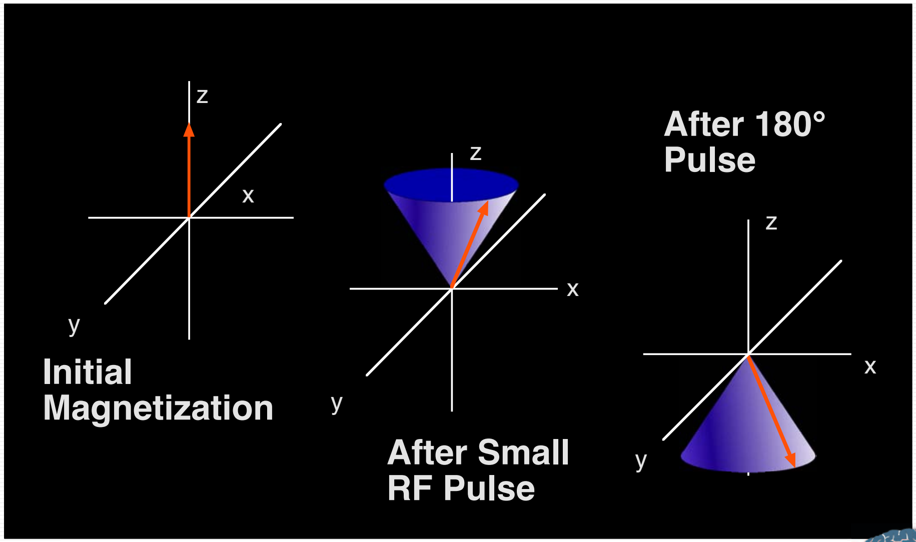
Small Flip Angles	Short	Long
Long	Proton Density	T2* Weighted
Short	Proton Density	T2* Weighted

Exercise in Afternoon Lab

Spin Echo Sequence



A 180° Pulse is not used in FLASH imaging



T2 and T2*

T2: Transverse Magnetization Decay from Spin-Spin Interactions

T2*: Transverse Magnetization Decay from Local Magnetic Field Variations

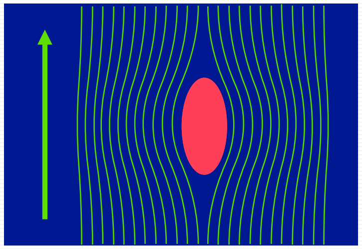
Magnetic Susceptibility

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

Magnetic Susceptibility

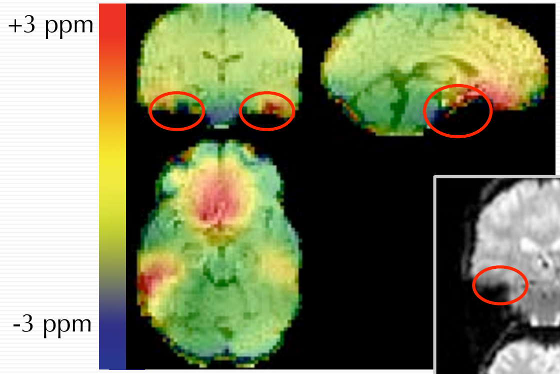
The Extent to Which a Substance Becomes "MAGNETIZED" when Placed Within a Magnetic Field

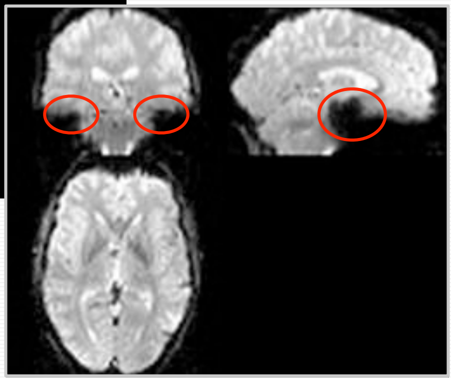
Applied Magnetic Field



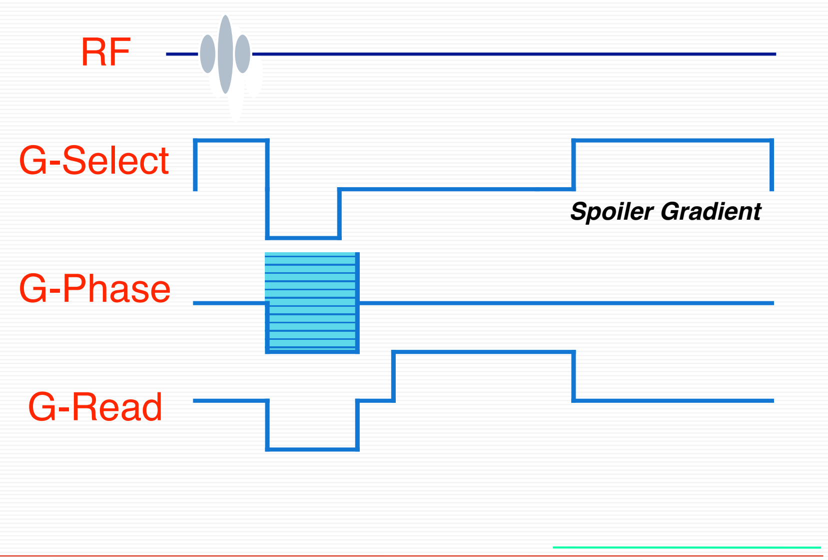
Objects with Susceptibility Different than Air Distort the Magnetic Field

Local field Variations Result in Signal Loss

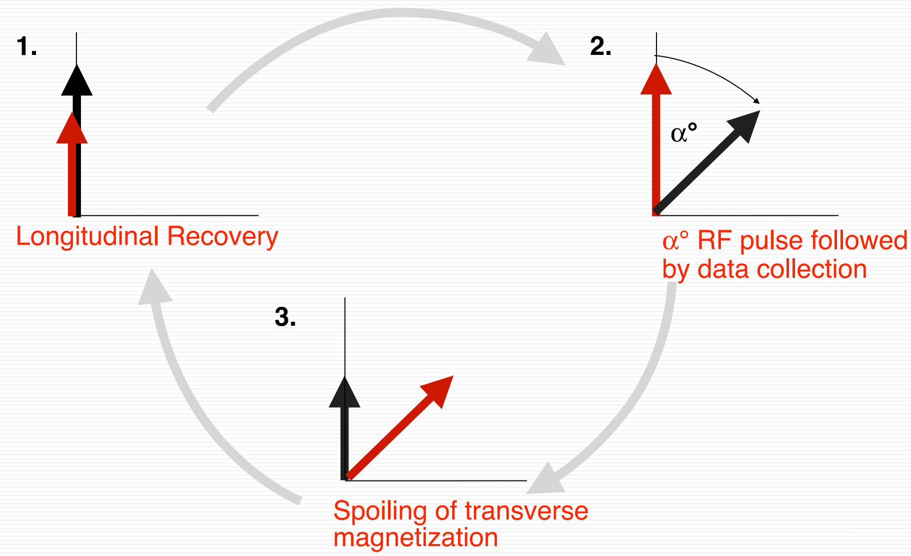




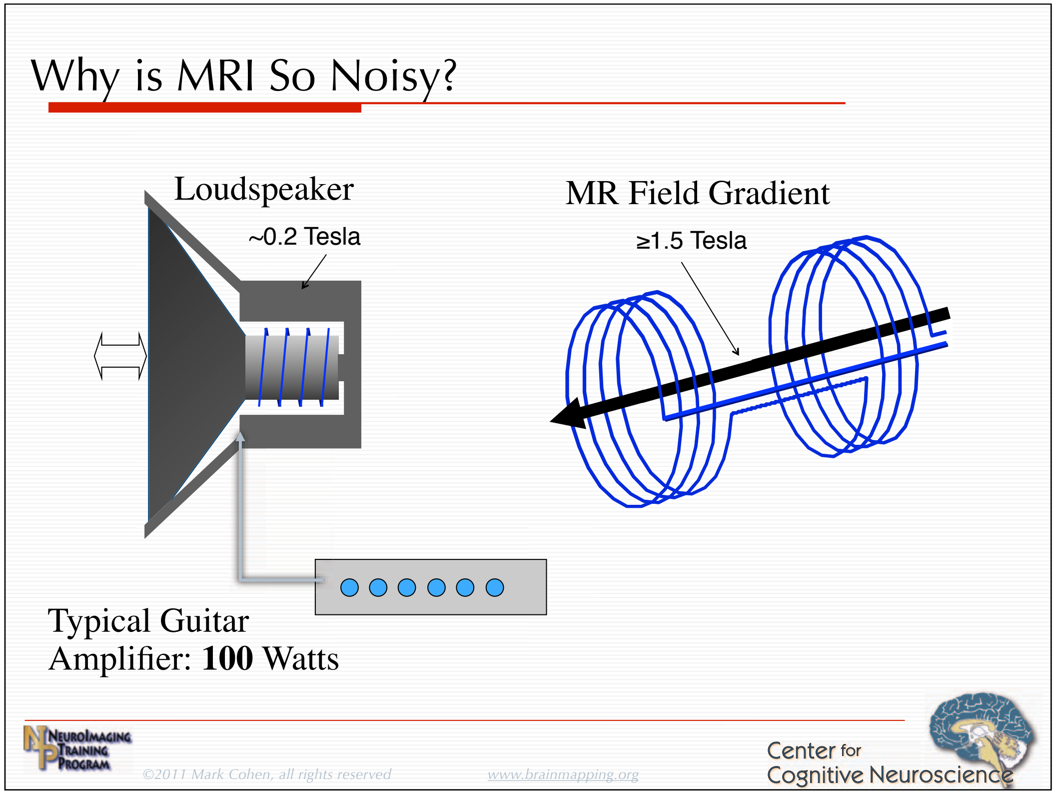
FLASH Timing Diagram



the FLASH Magnetization Cycle

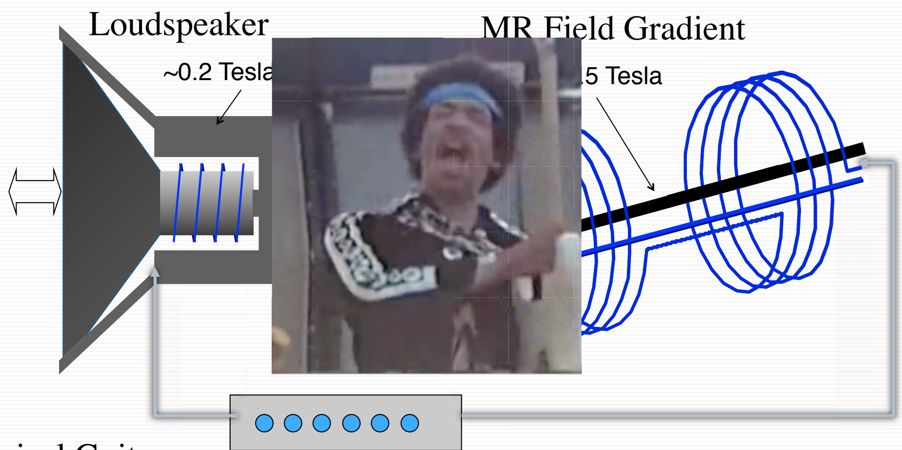


NeuroImaging Training Program



The construction of a gradient coil is very similar to the construction of a loudspeaker. Both utilize a coil of wire to produce a time-varying magnetic field in the presence of a large stationary field. The magnetic field in the coil experiences forces against the stationary field that tend to make it move. However, the magnitudes of all of the forces are much greater in MRI. Despite great efforts to reduce it, MRI is necessarily extremely noisy.

Why is MRI So Noisy?



Typical Guitar Amplifier: **100** Watts

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

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- Image Quality and Artifacts (48)

Brain "Activation" Leads to:

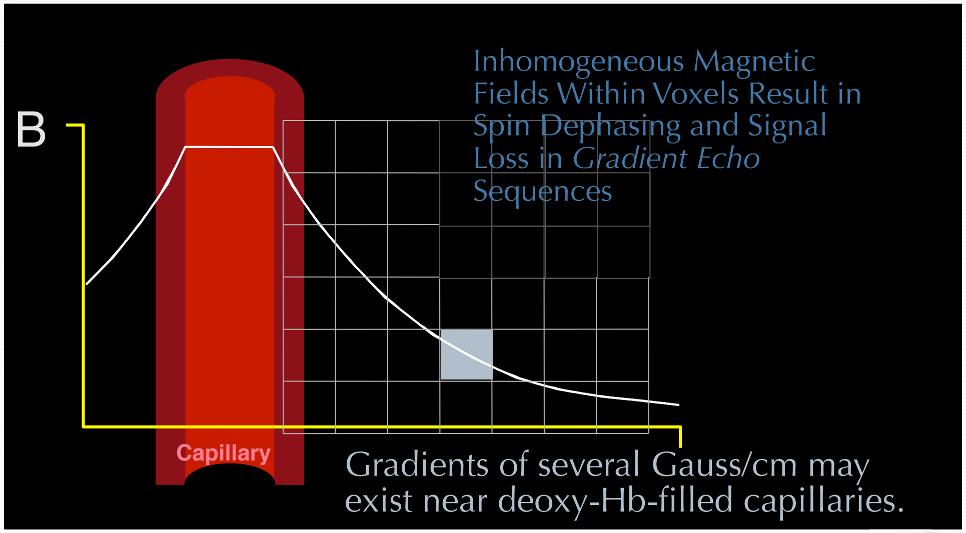
CBF	Increased	+ΔR1	
CBV	Increased	+ΔR2 (C+)	
O_2 Utilization	Increased slightly?		
Venous $[O_2]$	Increased	$-\Delta R2^*$	"BOLD"
Glucose Utilization	Increased	? Lactate	

R1=1/T1

R2=1/T2

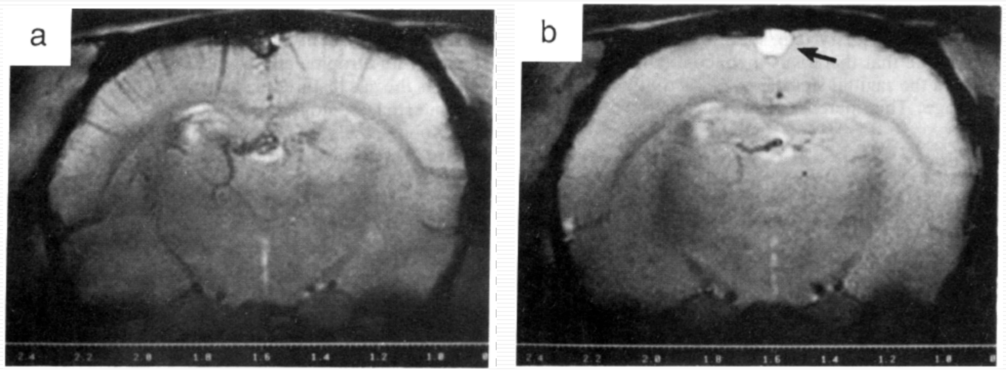
There are many biophysical effects associated with increased neuronal electrical activity. At least five coupling events are known to be observable during fMRI, though most studies at present rely on detecting the change in venous oxygen concentration which follows increases in synaptic activity.

Signal Losses from Spin Dephasing



MRI signal depends on the coherence of individual proton spins. In the presence of magnetic field distortions, the MR signal is reduced. Deoxygenated blood is paramagnetic and results in such field distortions in the immediate vicinity of the capillaries.

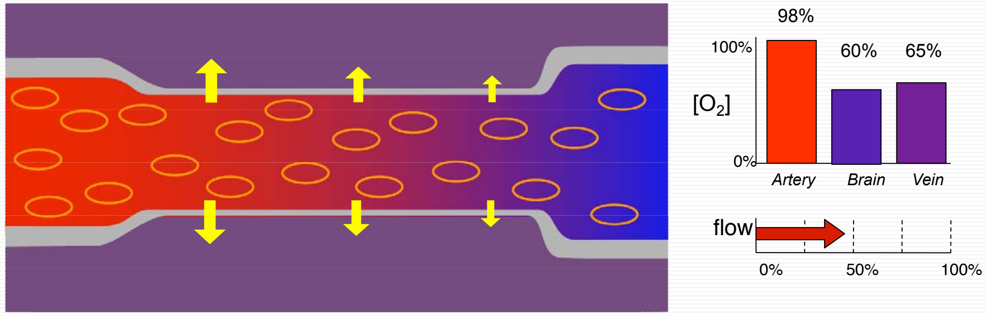
BOLD



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

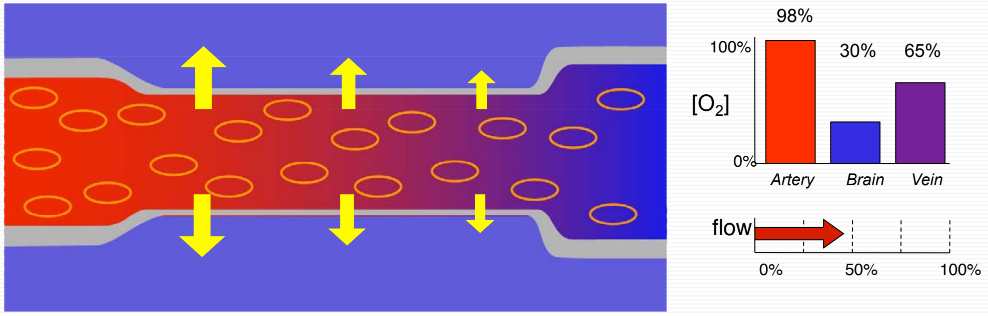
Why Does Venous O_2 Increase? (1)



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

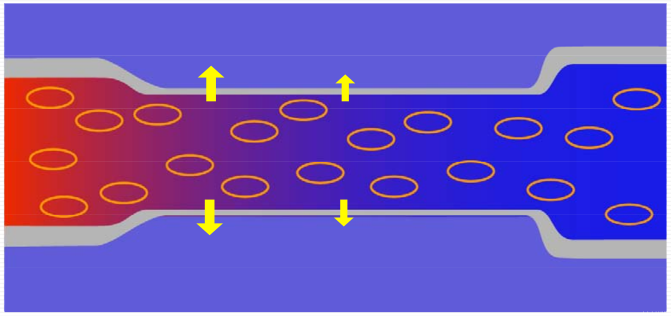
This series of slides outlines the dominant hypothesis as to why the BOLD signal is increased when brain metabolic demand is increased. Some believe, however, that this demand is driven not by oxygen but by glucose. The effects would be the same, however.

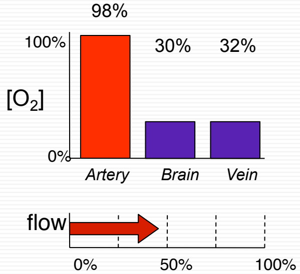
Why Does Venous O_2 Increase? (2)



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

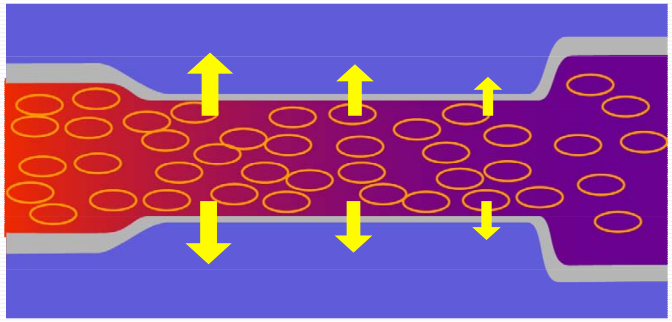
Why Does Venous O_2 Increase? (3)

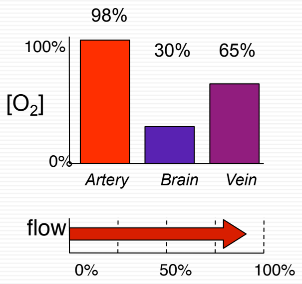




As oxygen flows across the capillary lumen it is depleted in the capillary and no further oxygen can be delivered

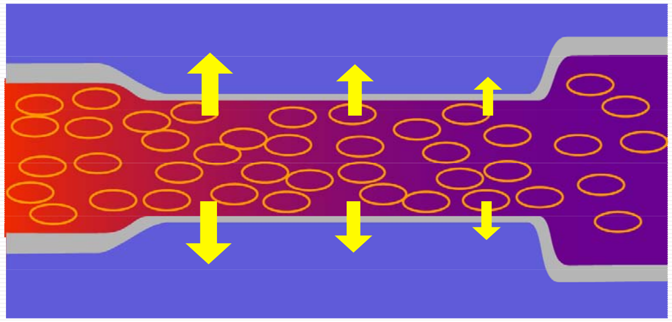
Why Does Venous O_2 Increase? (4)

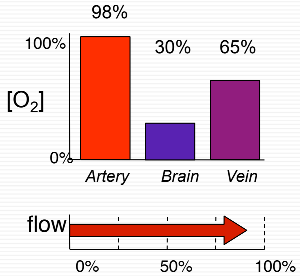




The vascular system responds by increasing blood flow so that more oxygenated blood is available throughout the capillary

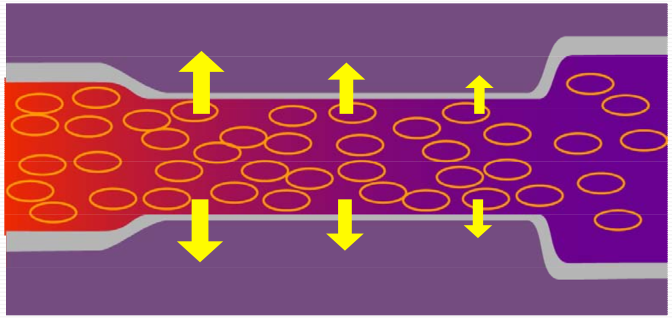
Why Does Venous O_2 Increase? (5)

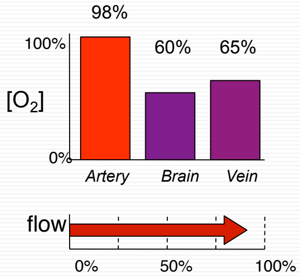




Because the blood flow is increased more oxygenated blood passes into the venous end of the capillary

Why Does Venous O_2 Increase? (6)





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

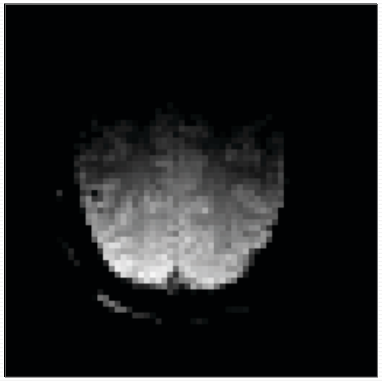
BOLD Contrast & Field Strength

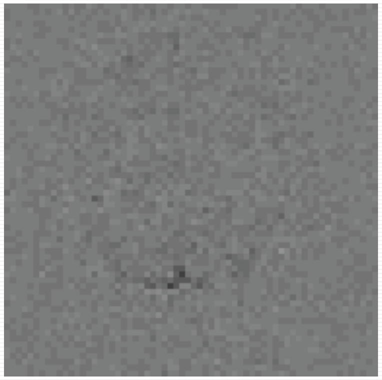
- BOLD Contrast arises from susceptibiity differences
- The absolute field distortion (from BOLD) is proportional to the magnetic field strength
- The absolute change in MRI signal is proportional to *both* the field distortion and the signal strength.

$$\boxed{\text{BOLD } \textit{should} \text{ go as } kB_0^2}$$

fMRI

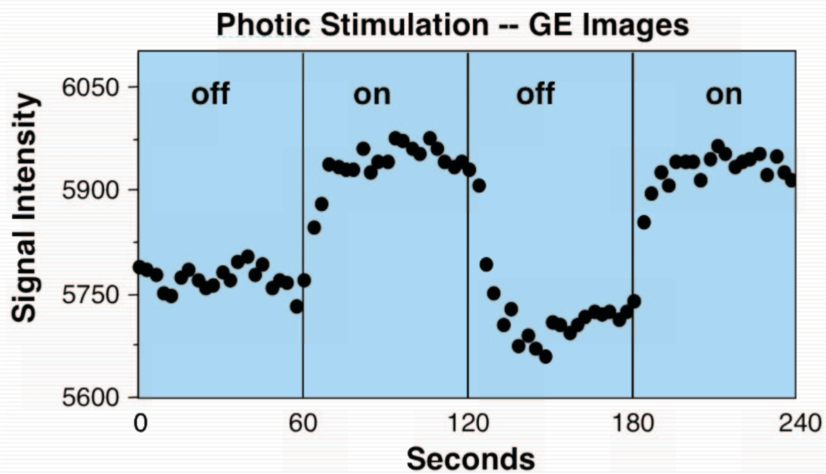
explores intensity variations in MR signal





intensity variations reflect venous [O2]

Gradient-Recalled Echo

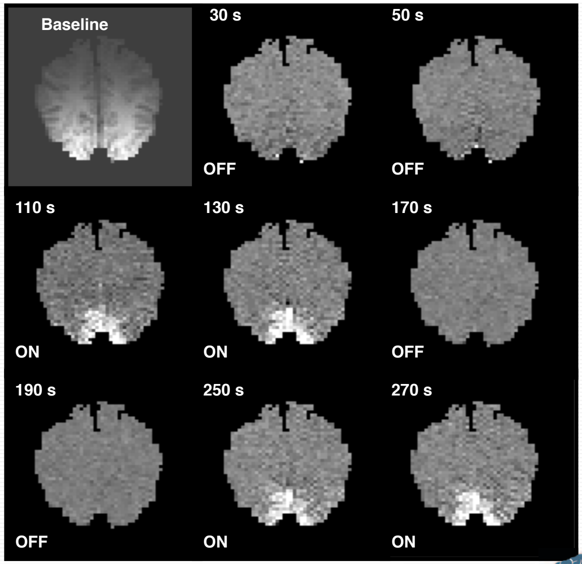


Ken Kwong

fMRI Signal increases relatively slowly (several seconds), and returns to baseline slowly.
No exogenous contrast needed

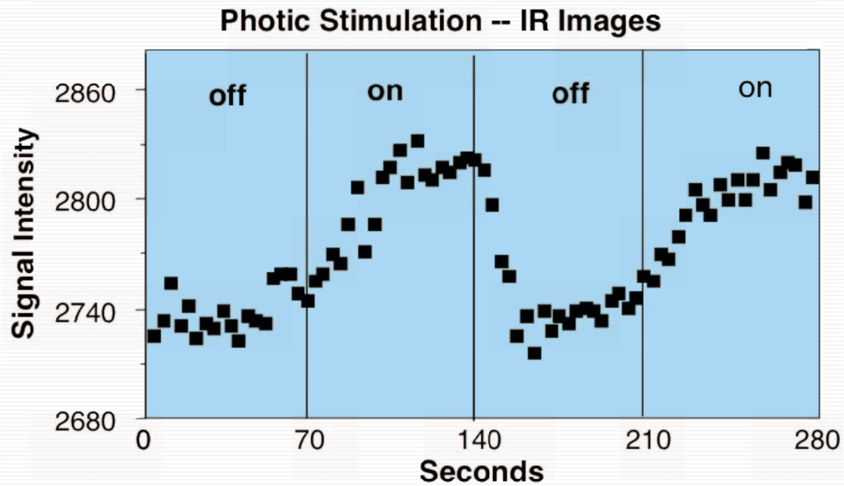
Ken Kwong

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



A second method is sensitive to blood flow changes, and demonstrates similar results.

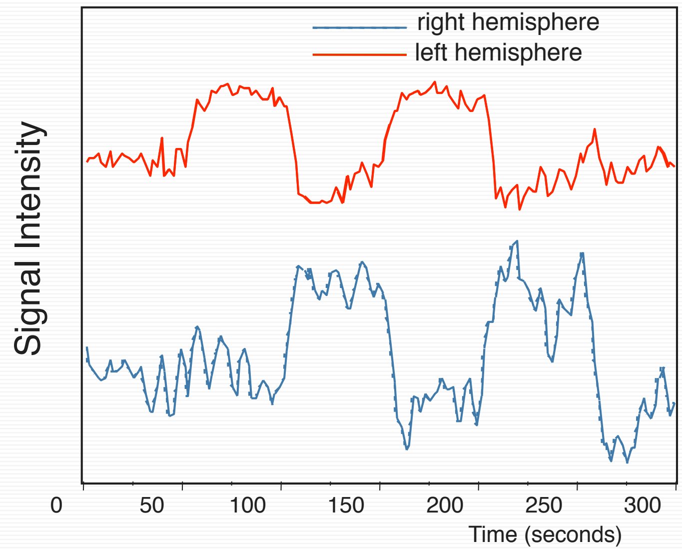
Inversion Recovery



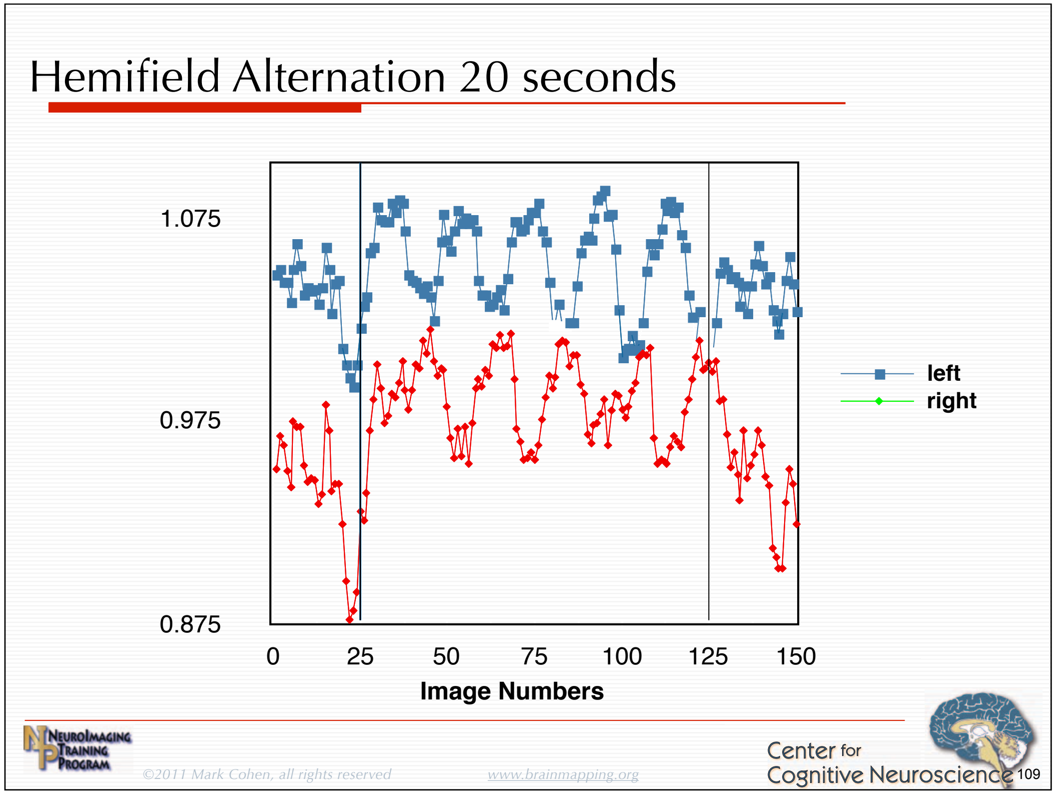
Ken Kwong

Note that the flow signal continues to increase for a longer time. This is due to complex effects of perfusion and proton exchange, as well as to possible vascular dynamic effects

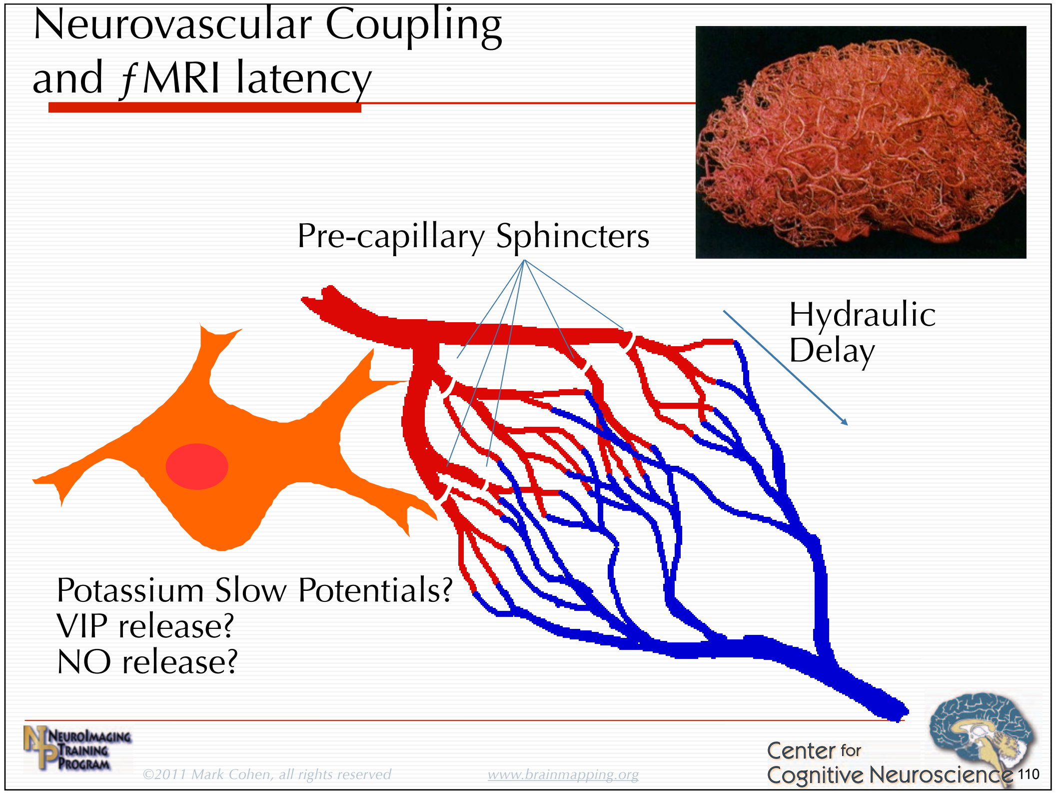
Hemifield Alternation



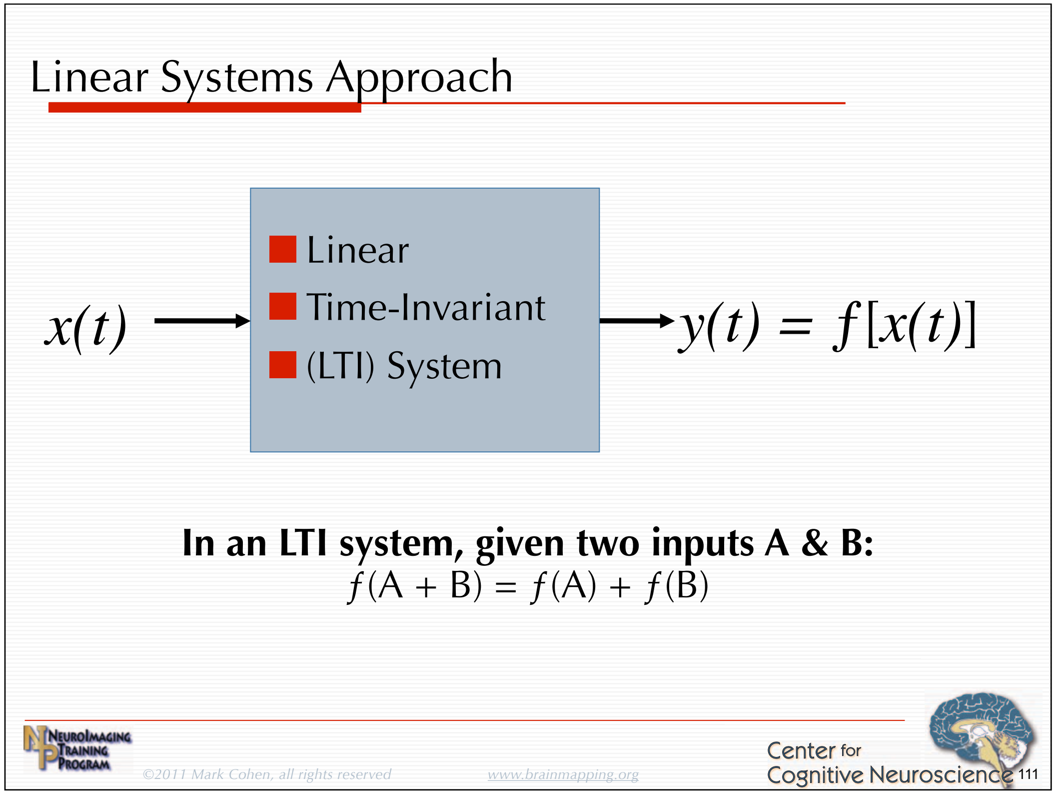
If stimuli are modulated relatively slowly, the BOLD fMRI response follows the stimulus timing rather faithfully.



When stimuli are varied quickly, the BOLD signal changes are much reduced and do not look a great deal like the stimulus timing.



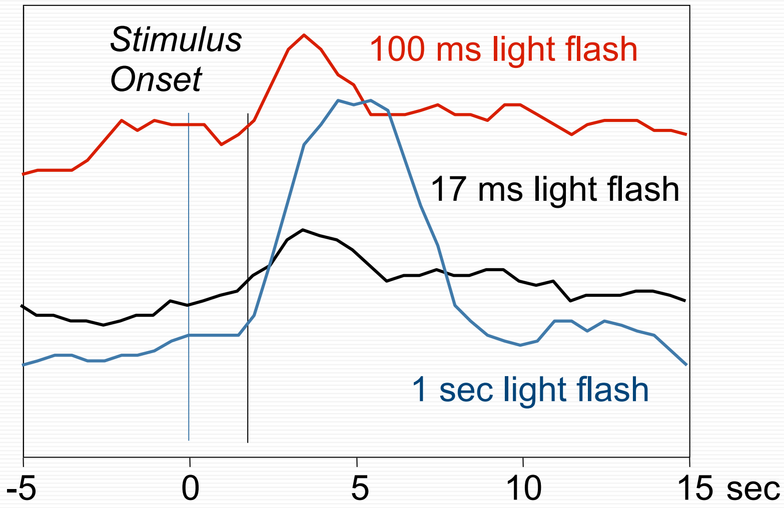
The fMRI response takes some time to occur. Much of this delay is thought to be related to the signaling from neurons to the vascular system. At this time it is not entirely clear how this signaling takes place, though there are many candidate signaling molecules and mechanisms that are probably all involved to varying degrees.



Almost all fMRI analysis is now performed under the rubric of linear systems analysis. This is because the properties of linear systems are very well-worked out and the toolsets available to analyze them are mature and relatively simple. A linear, time-invariant system is defined as one which obeys the property that the response of the system to two inputs presented together is the same as the sum of the responses to those two inputs presently individually.

One of the most important properties of a linear system is that by measuring its response to a standard input - usually an instantaneous transient input, we can calculate and predict its response to any input through the use convolution. Specifically, the predicted output, Y, to any stimulus, X is the convolution of X with the systems response, h, to an impulse: Y = X * h, where * is used to indicate convolution.

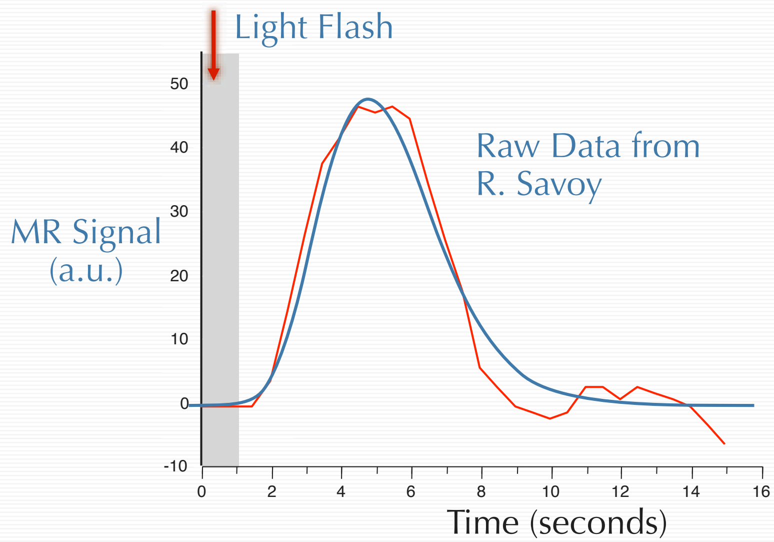
Response Latency vs. Stimulus Duration
Average of 10 recordings



Data courtesy of Robert Savoy

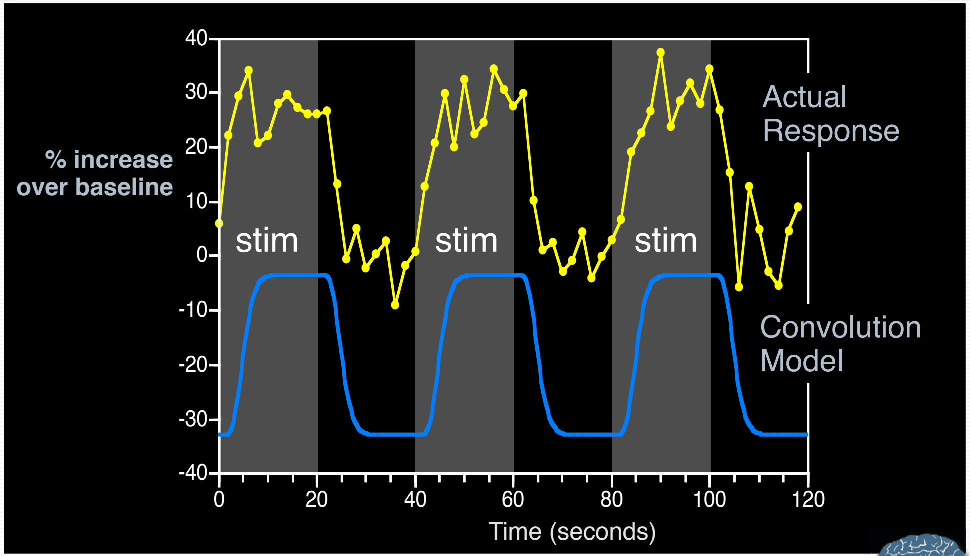
To capture the power of linear systems analysis for fMRI data it is useful to estimate the form of the brain's response to an impulse stimulus. These data of Robert Savoy show the response in V1 to brief flashes of light, which is a reasonable approximation to an impulse stimulus.

Brain Impulse Response



Using the observed V1 impulse response one popularly used model of the brain impulse response is the Gamma variate waveshape, shown here in blue. The Gamma function is a relatively function whose shape is very similar to the observed brain impulse data.

Convolution of Impulse Responses with Stimuli



The yellow points indicate the signal intensity observed in V1 in response to an intense flashing light stimulus that was turned on and off every twenty seconds. Convolving the time course of the stimulus with the Gamma function discussed above results in the estimate for brain activity shown in blue. Visual comparison of the two curves suggests that the convolution model is a reasonably good approximation to the observed response. The canonical means of detecting brain activation by fMRI is to model the brain response to the stimulus and to search for brain regions that behave similarly to the model. Thus, the convolution approach based on linear systems theory is a powerful tool for this work.

Linear Systems Approach

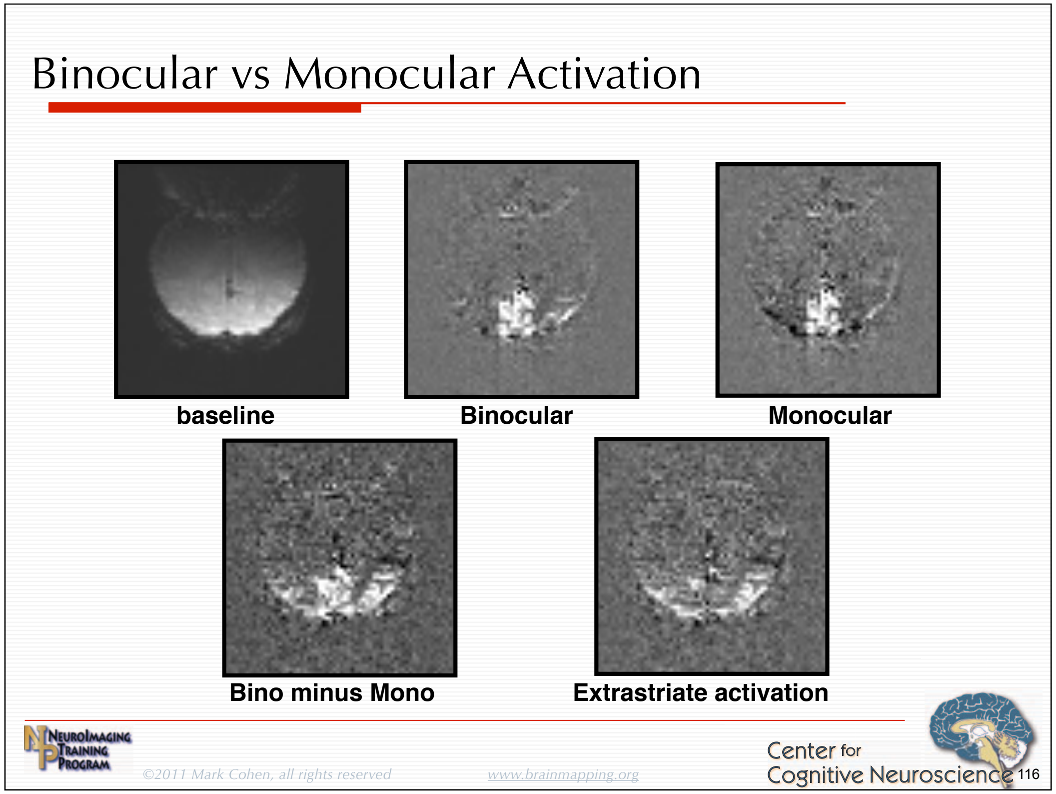
$x(t) \longrightarrow$

- Linear
- Time-Invariant
- (LTI) System

$\longrightarrow y(t) = f[x(t)]$

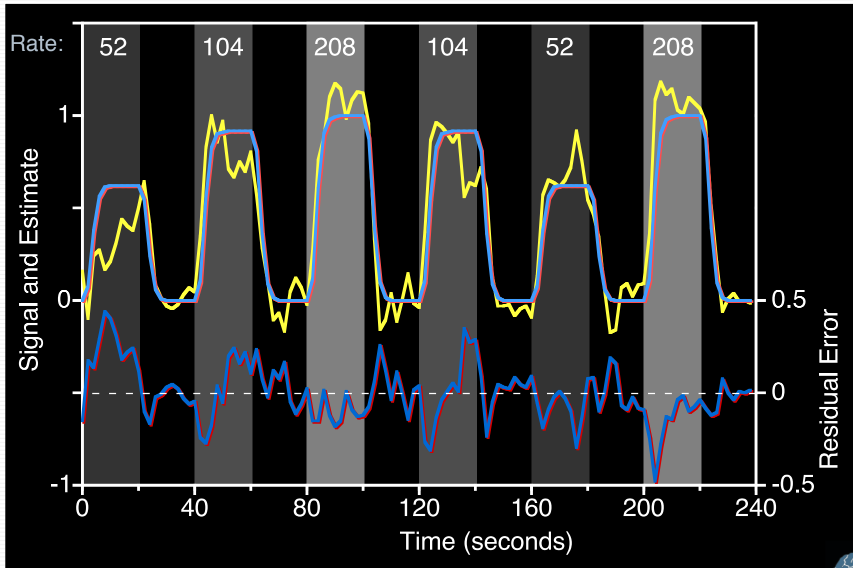
In an LTI system, given two inputs A & B:

$f(\mathrm{A} + \mathrm{B}) = f(\mathrm{A}) + f(\mathrm{B})$



Given the importance of the linear systems analysis, it is instructive to test whether the brain fMRI signal response is linear at all. These data from Ken Kwong show the brain responses to visual stimulation to both eyes simultaneously and to one eye alone. Because the primary visual cortex is known to be independent for each eye, we would expect the difference between the binocular and monocular stimulation to be the same as the response to the other eye alone. The data however show that this is simply not the case. Thus our assumption of linearity is strongly violated.

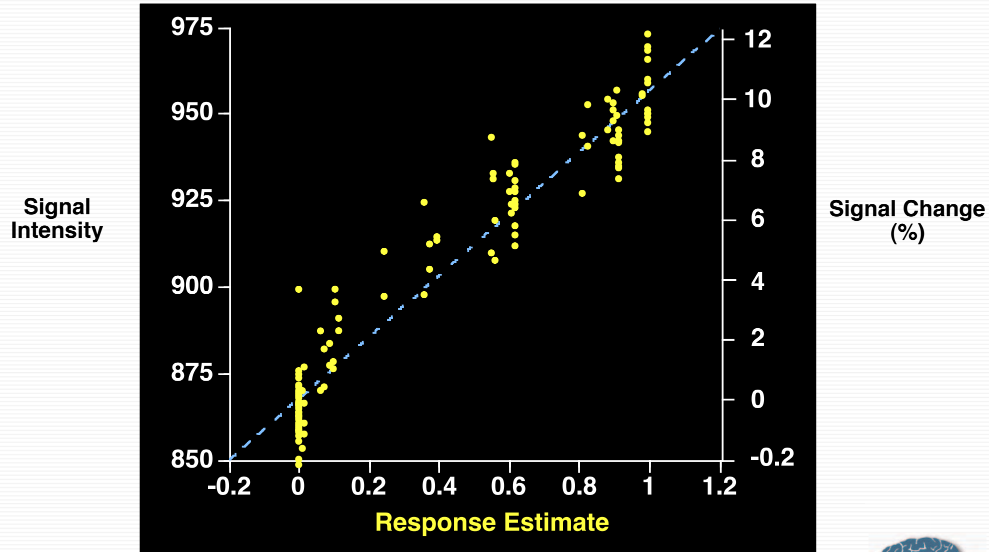
Amplitude-weighted Linear Estimate



Various degrees of non-linearity can be incorporated into the linear systems analysis in a principled manner. Here, the estimated BOLD response was modeled as increasing in amplitude with the log of the rate at which subjects performed a finger motion task. Notice that the response estimates closely fit the data.

People using fMRI for their data collections should be cognizant of the assumptions made in the analysis and thoughtful of how they relate to the underlying physiology of the brain. In many cases, the assumptions of linearity will produce grossly distorted activation maps.

Estimated vs. Actual ƒMRI Response



These are the data from the prior slide plotted instead as the observed signal intensity as against the modeled intensity and show that the log transform model closely fits the actual data.

The Plan

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Seventh Inning Stretch

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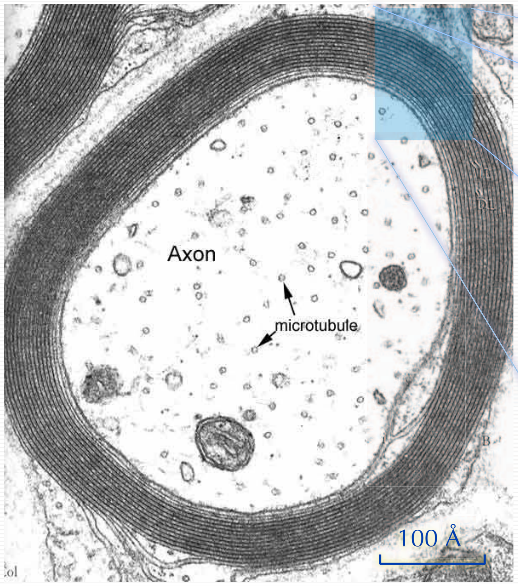
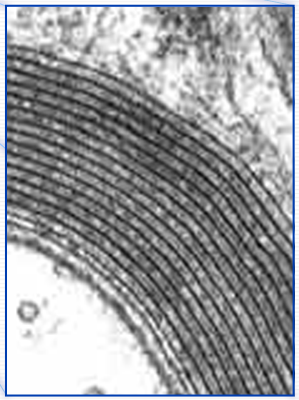
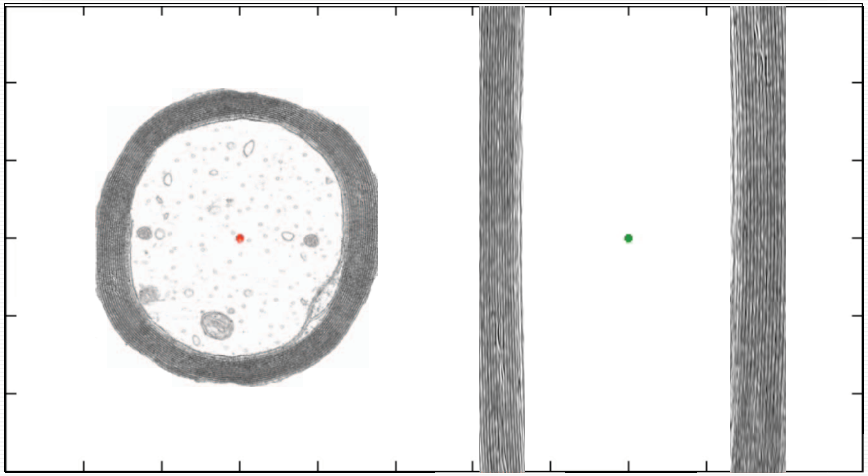
- Diffusion and Summary (9)
- Image Quality and Artifacts (48)

Myelin Sheath



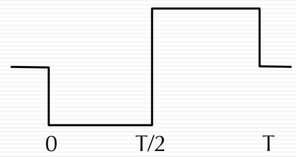


Isotropic vs. Anisotropic Diffusion



Motion Effects on Phase

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

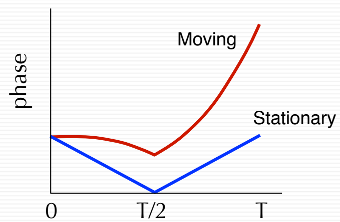


$$x(t) = x_0 + vt + \frac{at^2}{2} + \dots$$

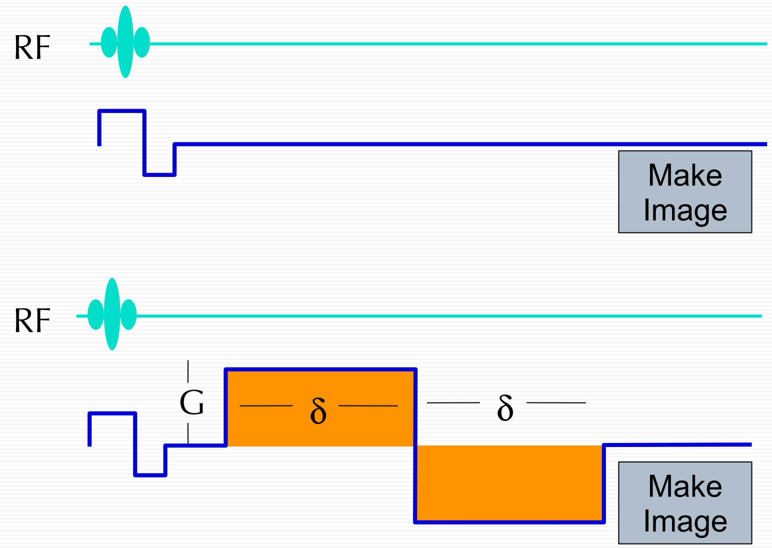
$$\varphi = \gamma \int_0^T G(t)x(t)dt$$

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

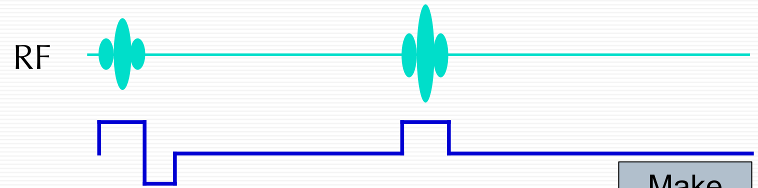
$$= \gamma \left[\frac{vT^2}{4} + \frac{23aT^3}{24} + \dots \right]$$



Diffusion Gradients and Signal

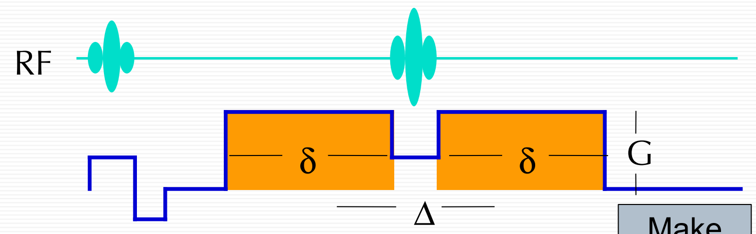


Diffusion Gradients and Signal



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

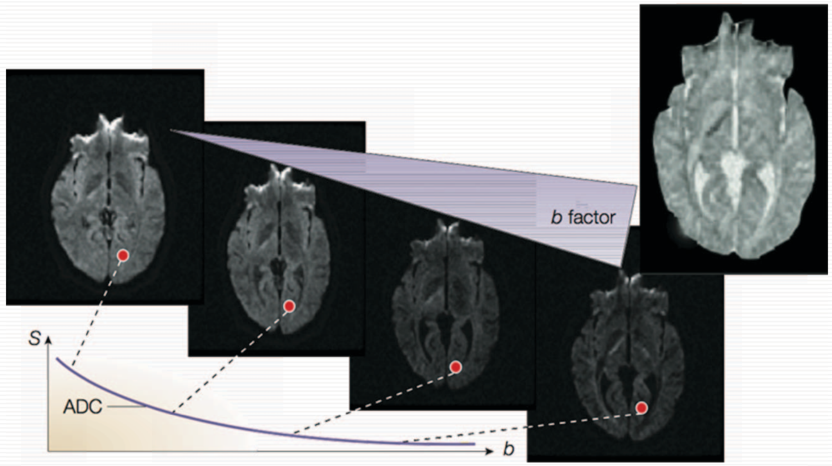
Make Image



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

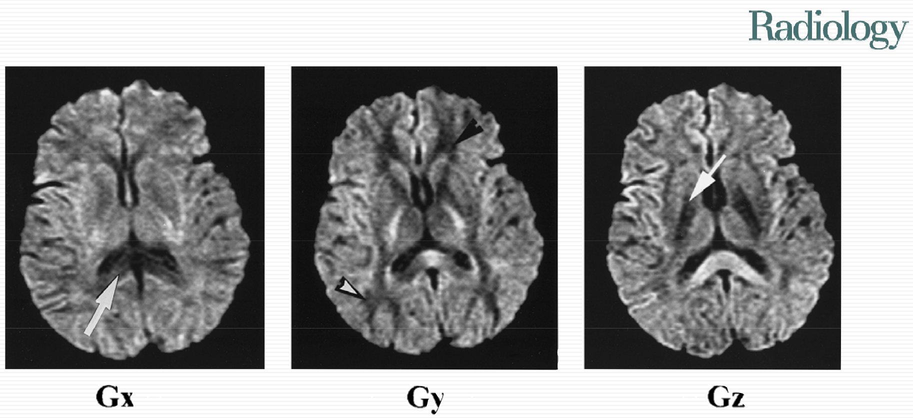
Make Image

Diffusion Attenuates MR Signal



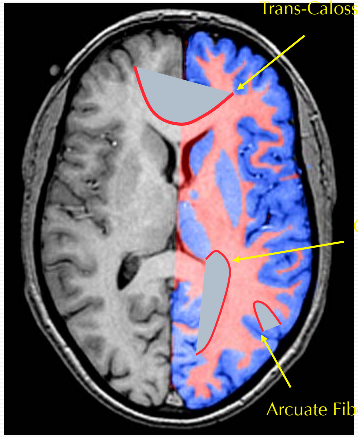
Denis Le Bihan, Nature Reviews in Neuroscience 4:469, 2003

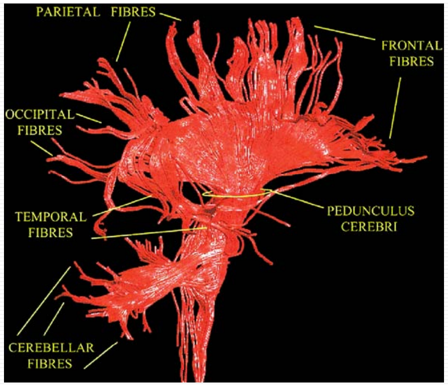
Brain Diffusion is Anisotropic



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

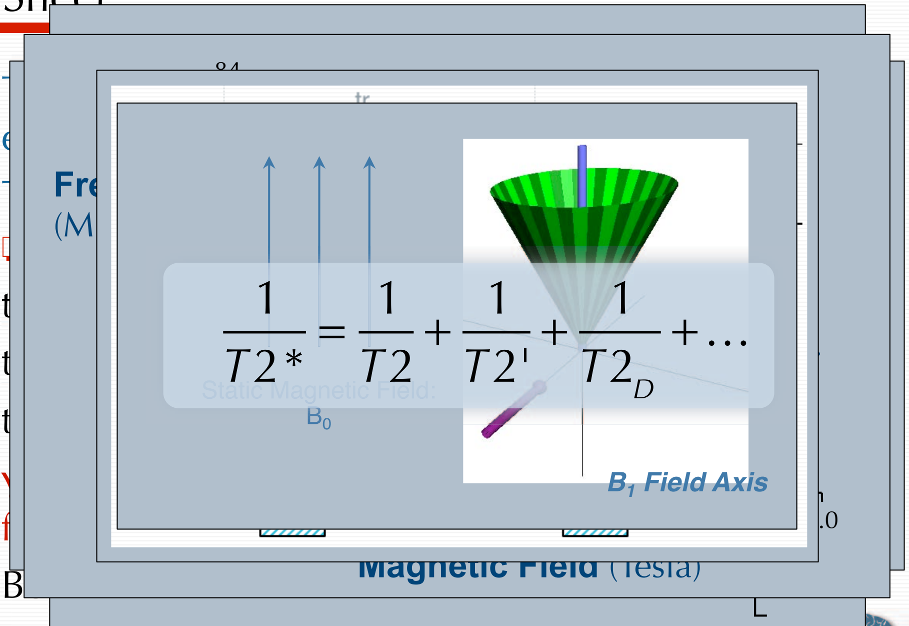
White and Gray Matter





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

Crib Sheet



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

- B1 - Rotating Magnetic Field

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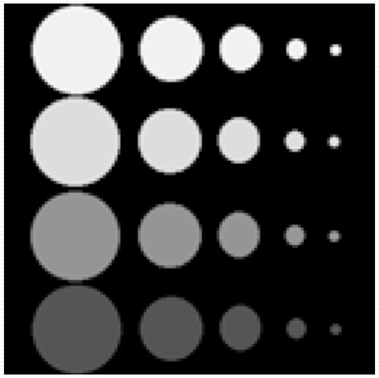
An “Equation” in Resolution

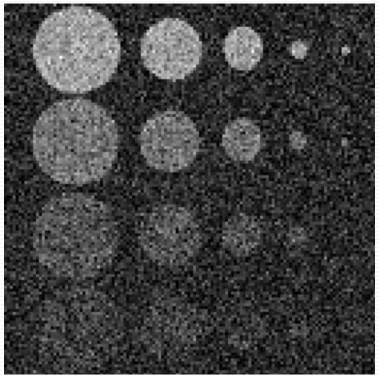
Because MR is an emission modality the temporal resolution, spatial resolution and contrast are inter-dependent:

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

where B_0 is the field strength.

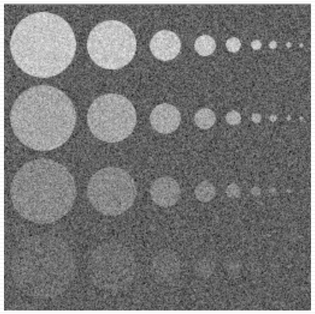
Contrast to Noise Ratio



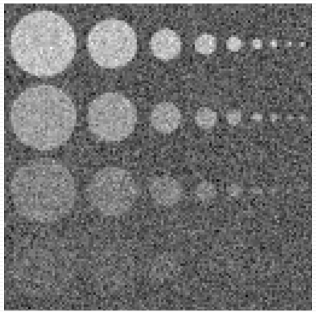


CNR vs. Resolution

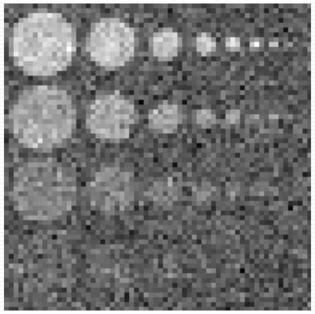
Signal/Noise Ratio Held Constant



Imaging time = 16X

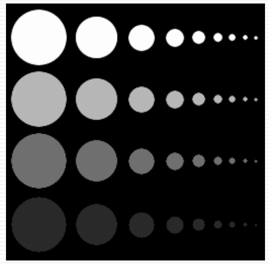


Imaging time = 4X



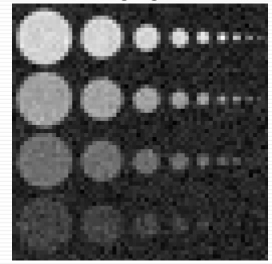
Imaging time = 1X

CNR vs. Resolution



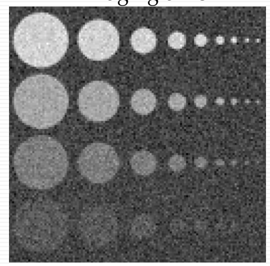
Noise free

Imaging time = 1X



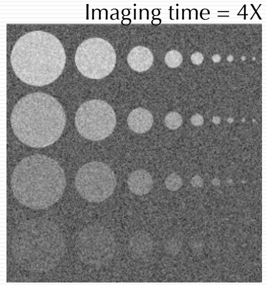
64 X 64

Imaging time = 2X



128 X 128

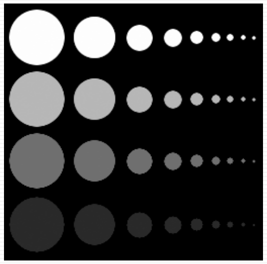
Imaging time = 4X



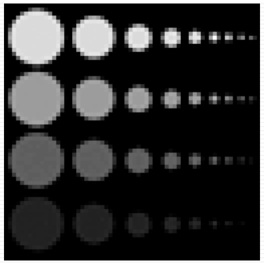
256 X 256

Minimum Imaging Time

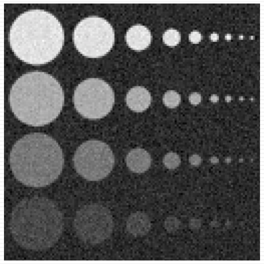
CNR vs. Resolution



Noise free

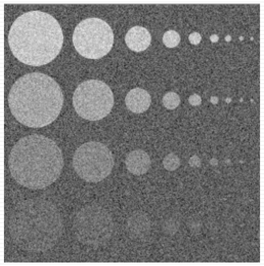


64 X 64



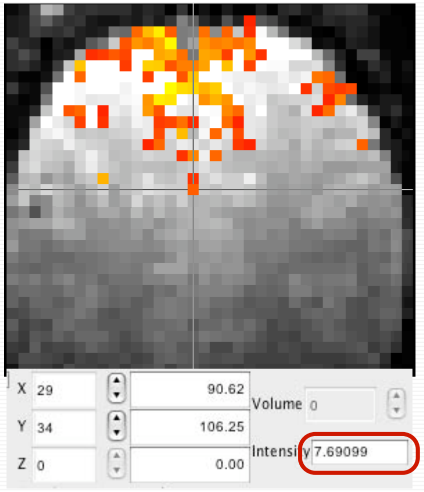
128 X 128

Imaging Time Held Constant

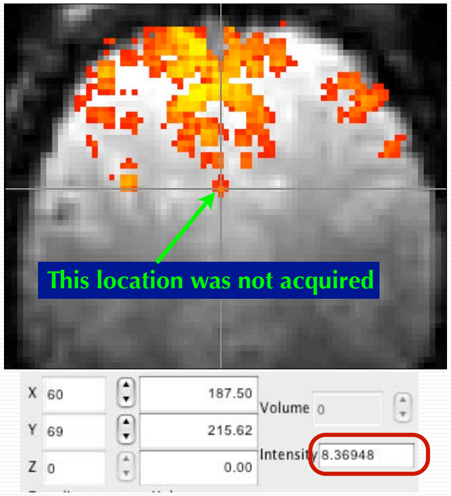


256 X 256

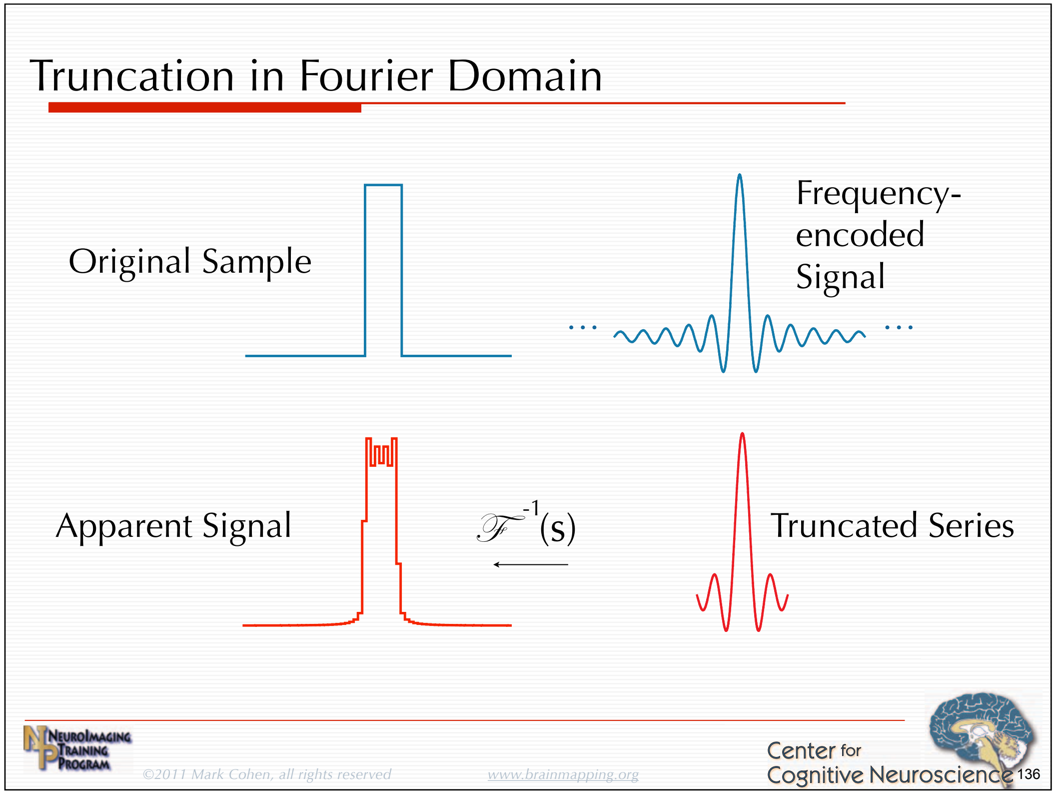
Interpolation



Native Resolution

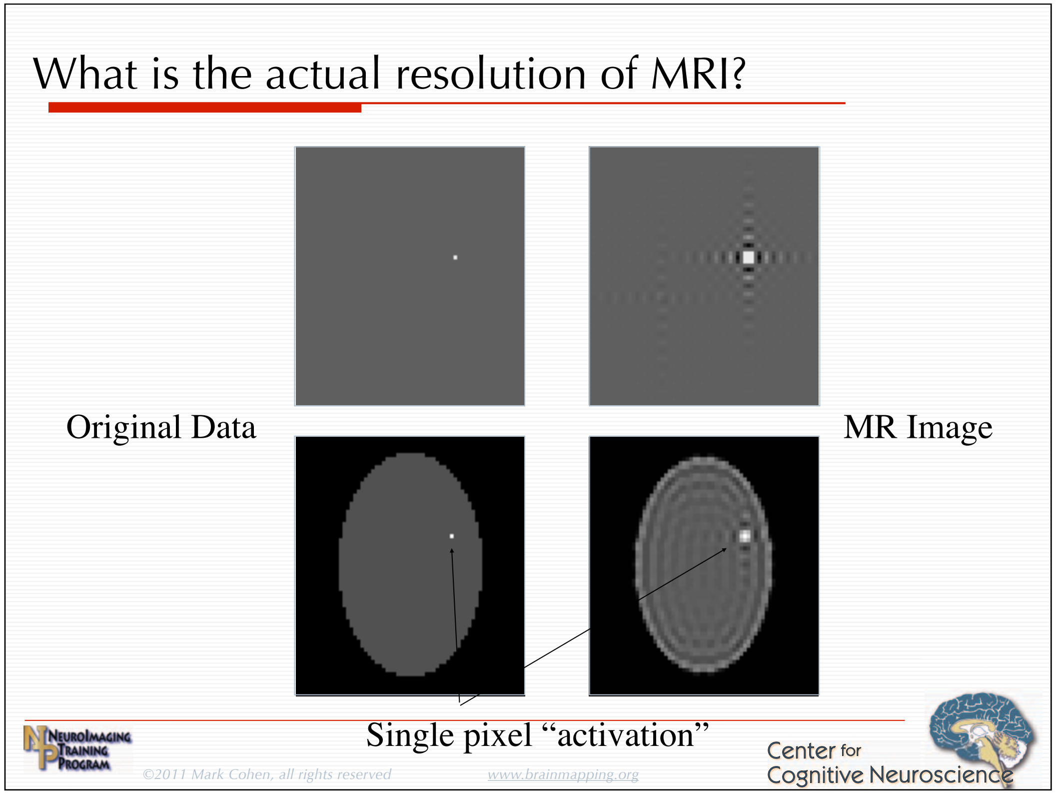


Bilinear Interpolation



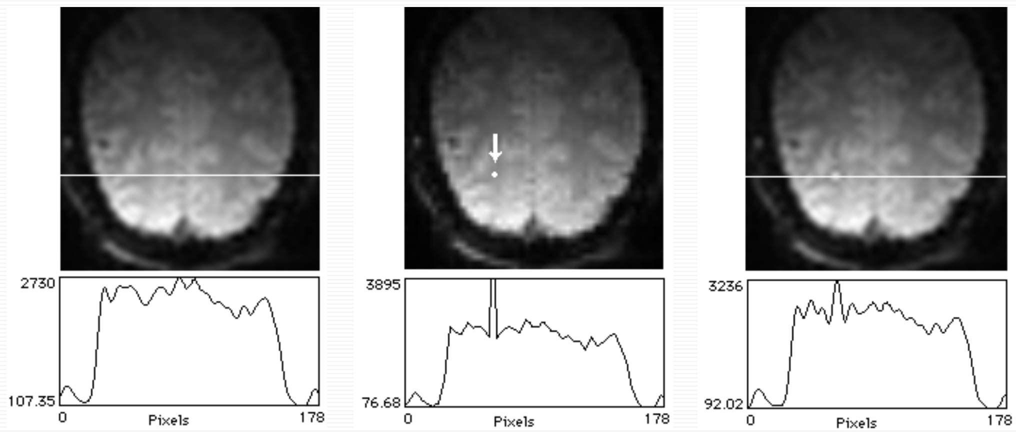
Remembering that the MRI raw data encode position by frequency, it is useful to consider factors that control the fidelity of that encoding. The Fourier transform represents the amplitudes (and phases) of the frequencies in a signal or waveform. Consider a simple rectangular signal as shown at top left. Its representation in frequency is shown at top right. Notably, the transform has infinite tails because the waveform contains energy at all frequencies.

If the waveform at upper right were the MR signal for a rectangular object, we would necessarily be able to acquire only part of it as doing otherwise wold require infinite time. When we then take the Fourier transform of the truncated signal, we find a distorted representation of the original waveform. Notably, the signal now contains ringing at the edges.



Because actual MR imaging is always based on a truncated signal collection, the resulting images are always distorted. In this simulation, we consider the images that would result from a small area of bright signal. The image representation will spread far away from the signal source as a sequence of dark and light bands. Further, a distinct, though low intensity, focus of signal will appear in the opposite X and Y locations on the image. If the bright location were a focus of brain activation that caused a BOLD signal increase, the apparent location of the activity would be represented throughout the image. An accurate statistical map based on the images would include signal decreases in the pixels next to the center of the activation.

The Actual Resolution of ƒMRI



http://porkpie.loni.ucla.edu/BMD_HTML/SharedCode/MRArtifacts/MRArtifacts.html

The apparent representation on an MR image of a signal point of true signal increase

Bandwidth and Readout

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

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

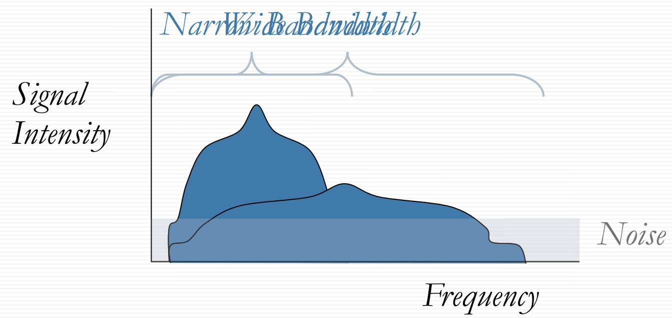
- Bandwidth decreases with readout duration:

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

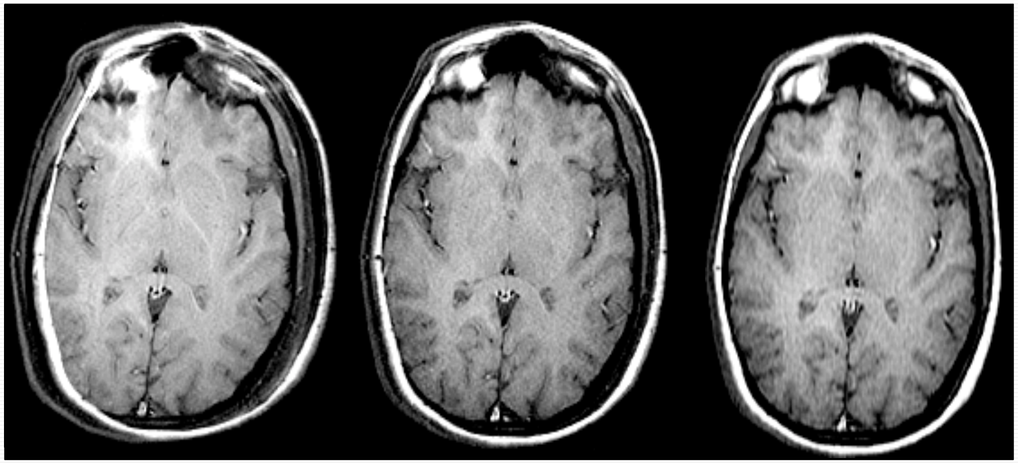
Bandwidth and SNR

Decreasing the Bandwidth Improves SNR:

Imaging Time is INCREASED and high frequency noise is excluded



Bandwidth

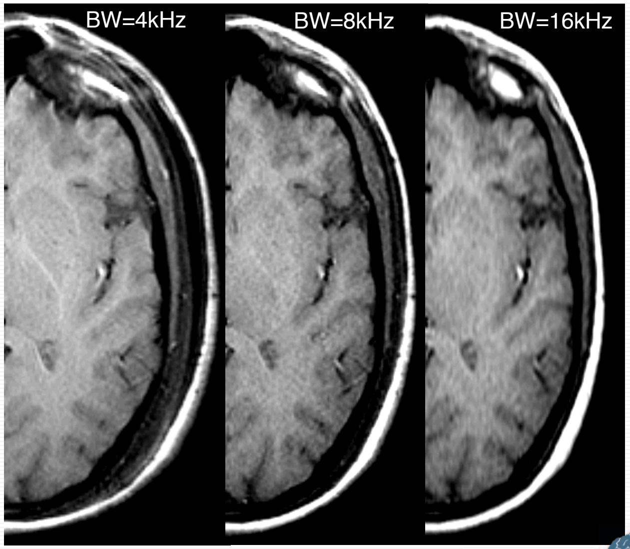


BW=4kHz BW=8kHz BW=16kHz

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

Bandwidth

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

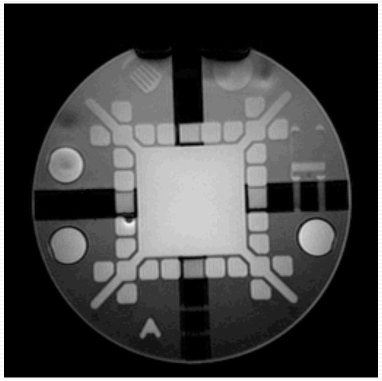


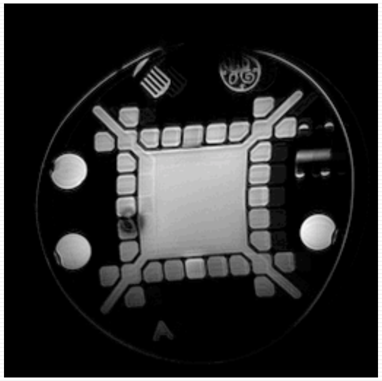
NeuroImaging Training Program

www.brainmapping.org

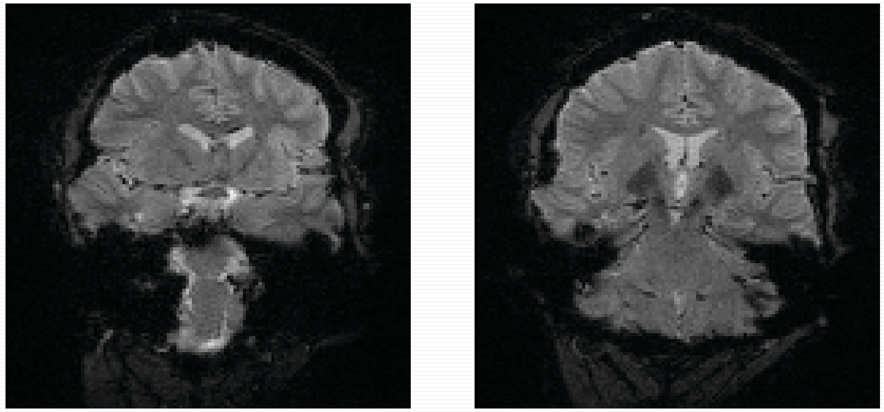
Center for Cognitive Neuroscience

Shape and Bandwidth





Distortions are More Severe at High Magnetic Field Strength

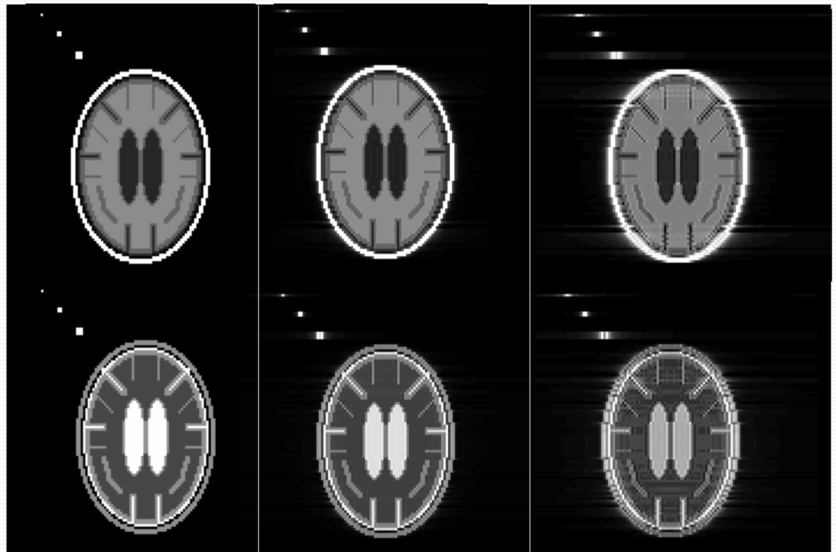


Variation in sample magnetization of is proportional to field strength.

High Field images lose more signal from field inhomogeneity

Location is detected by the frequency of spin precession, which in turn is determined by the local strength of the magnetic field. If the field is distorted, the images will be as well. Unfortunately, the subject him or herself distorts the primary field because body tissues magnetize unequally. The images become distorted by this. Various pulse sequence factors either tend to mitigate or amplify this effect, but the combination of parameters used to create fMRI images tends to amplify these shape distortions.

Apodization from Long Readouts



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

Exercise in Afternoon Lab

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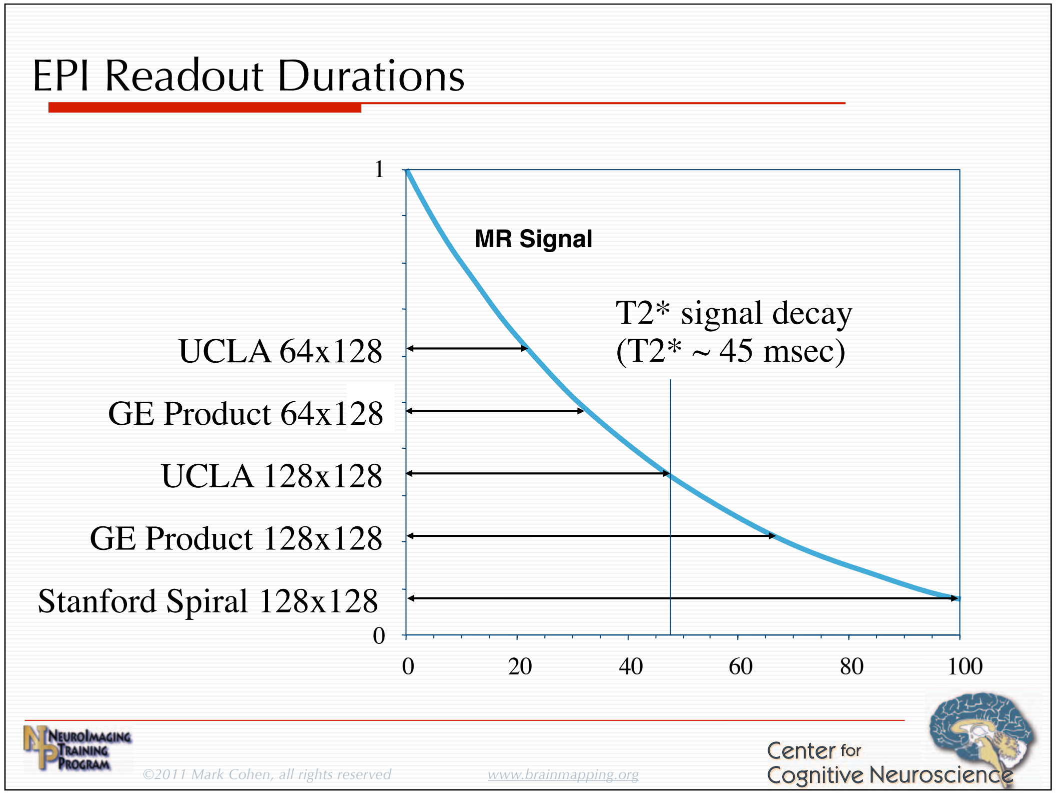
If the readout period is long compared to T2*, the images tend to become blurred or apodized. This simulation shows the blurring that would occur with various readout durations. T2* in the human brain is typically around 40 msec (but depends strongly on the scanner itself) and readout periods of 50 to 80 msec are not uncommon. Thus the images are detectably blurred. This simulation considers only rectilinear k-space traversals, such as EPI. In spiral scans the blurring that results is more circular and much more complex.



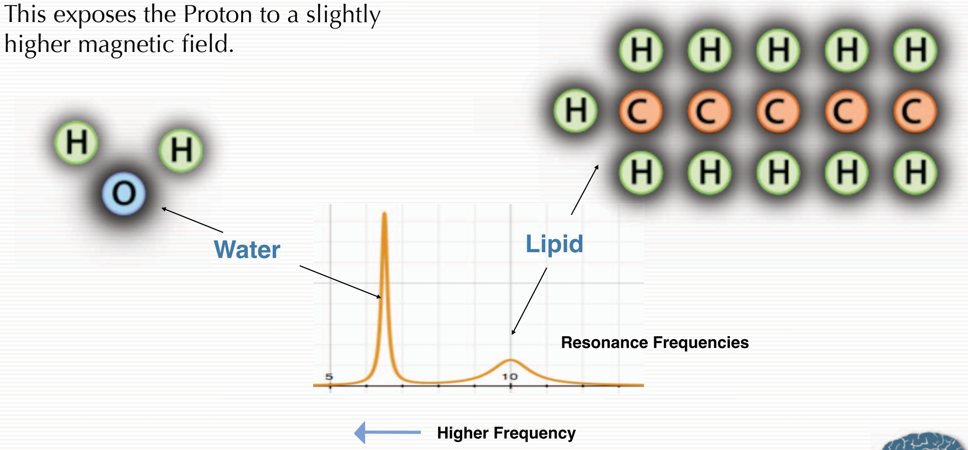
Henceforth all of the discussion about gradient encoding and image distortion pointedly neglected T1 and T2 effects on the signal. The actual effects on the spatial localization are not insignificant, however. During the time that we are encoding the location, the signal is changing. For example, during the readout of an echo-planar or spiral scan, T2* effects cause the signal to decrease. In long readout sequences, such as spiral scans the signal may decrease by 80% during the actual readout.

The Origin of Chemical Shift

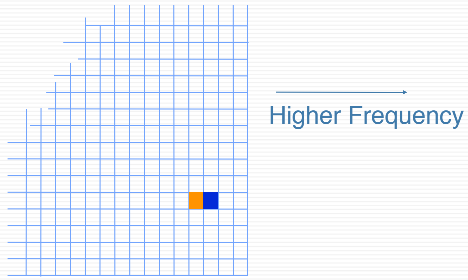
In water, electrons move from Hydrogen towards Oxygen.

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

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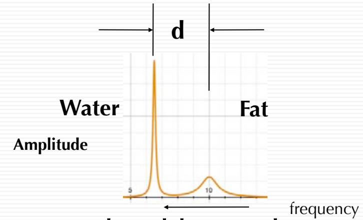
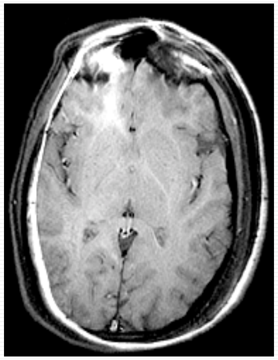
Chemical Shift

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

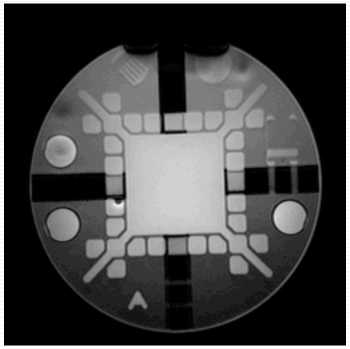
Which is:	*with a 32 kHz readout*
75 Hz @ 0.5 Tesla	< 1 pixel
150 Hz @ 1.0 Tesla	≈ 1 pixel
220 Hz @ 1.5 Tesla	> 1 pixel
440 Hz @ 3.0 Tesla	≈ 3.5 pixels

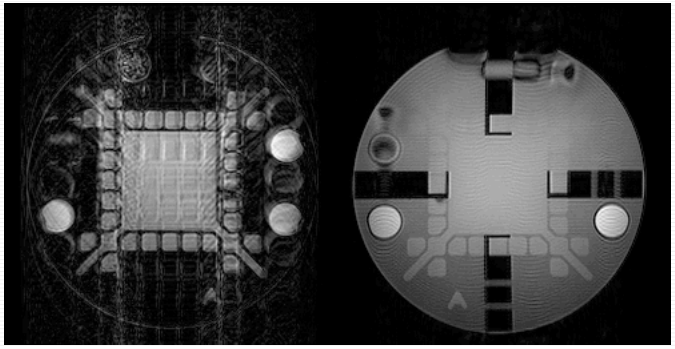




Lowering the Bandwidth/pixel increases the Chemical Shift in pixels

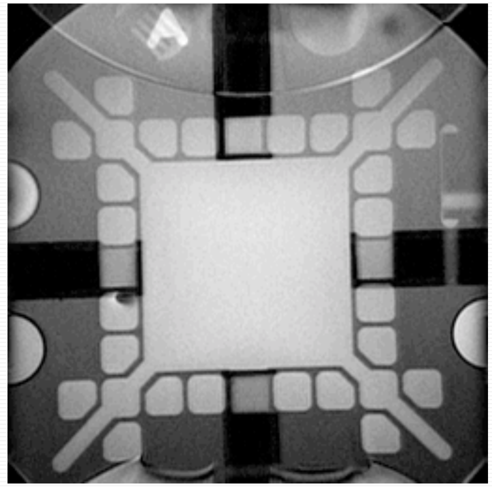
Motion Artifact



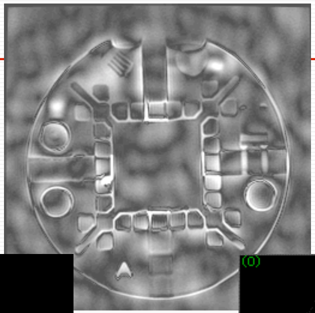


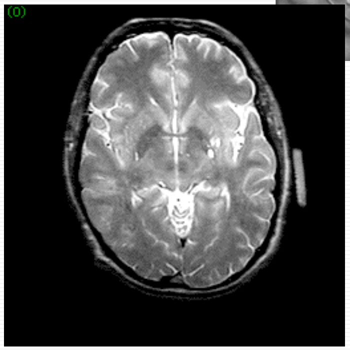
http://airto.ccn.ucla.edu/BMCweb/SharedCode/MRArtifacts/MRArtifacts.html

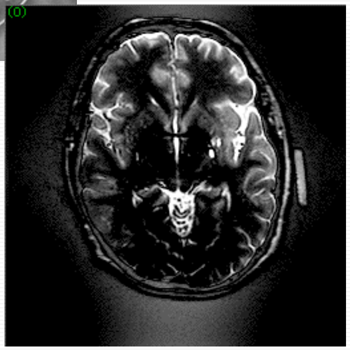
Aliasing



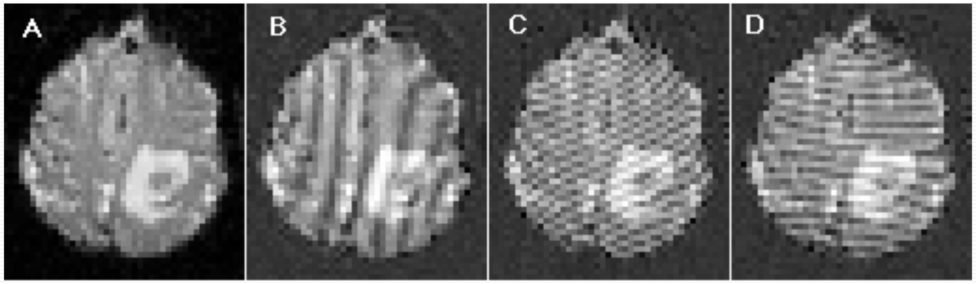
Saturation



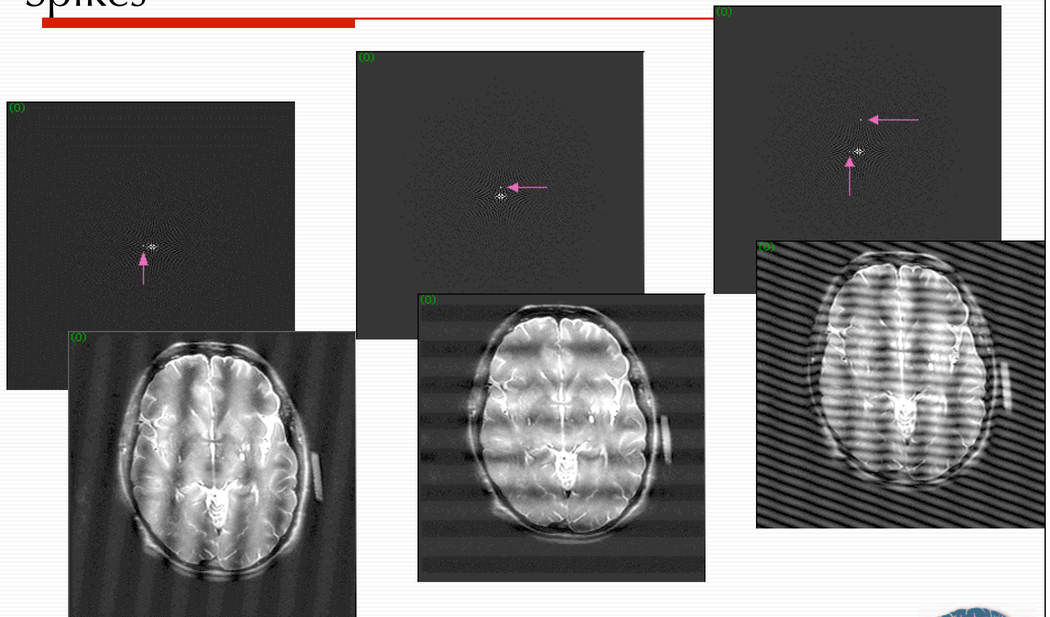




Spikes



Spikes



Exercise in Afternoon Lab

Image Quality

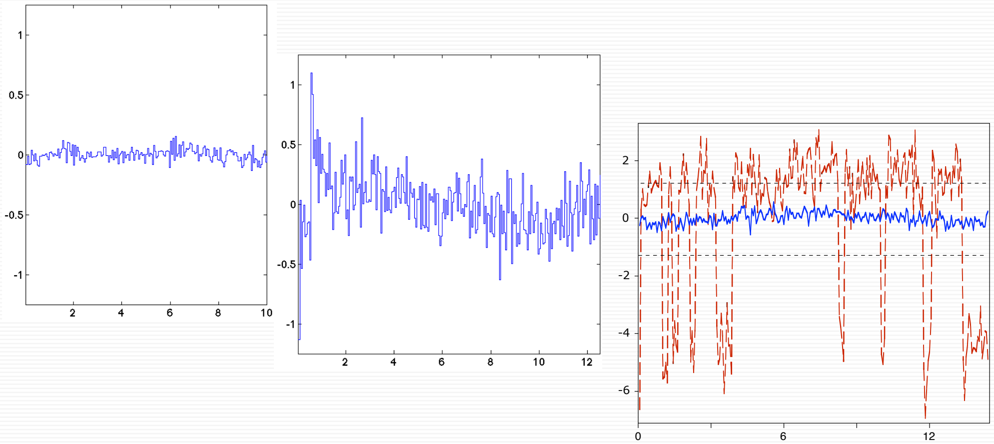
- SNR is Very Limited in MRI
- Feature Detection Falls Rapidly with Loss in Contrast to Noise Ratio
- Usable Resolution is NOT the Same as Voxel Size
- Spatial Encoding Artifacts in MRI May Have Complex Appearance
- Edge Ringing and Blurring are Related to Parameters Such as Contrast

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

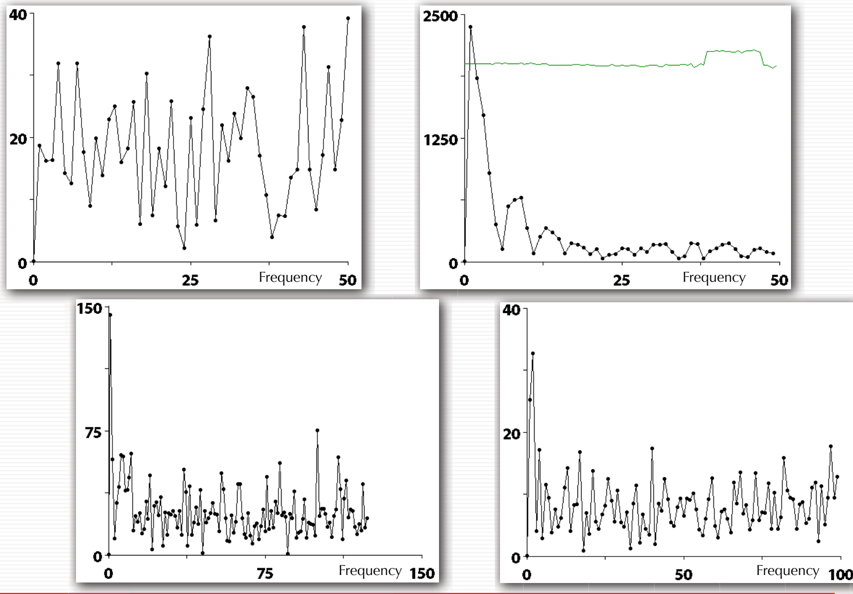
Instrument Variation

- 1. System Instability

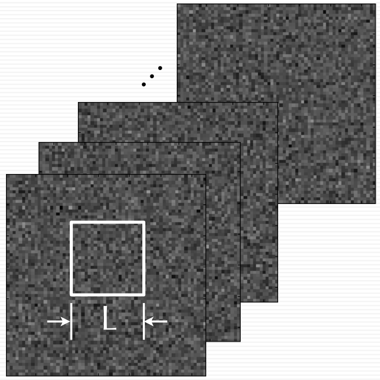


Mean Intensity Variation

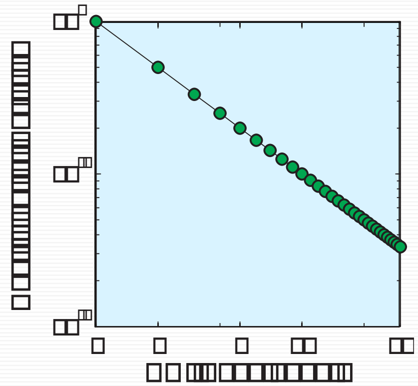
Scanner Autocorrelation



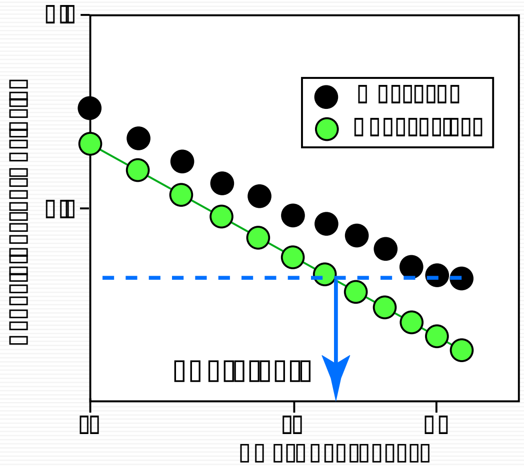
the Weisskoff Plot



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

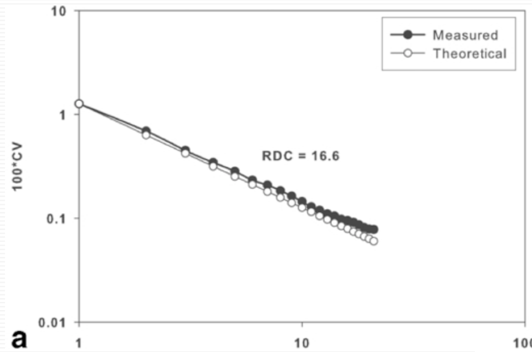


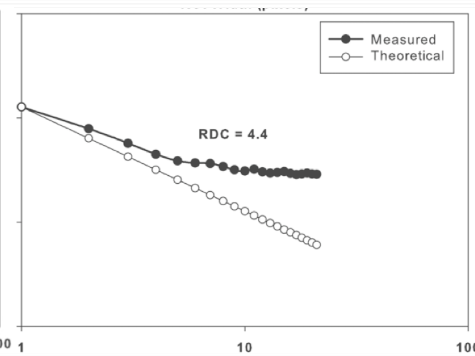
Deviations from the Theoretical Curve are Evidence of Correlated Noise

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

Weisskoff Plot

Weisskoff R. Magn Reson Med 36:643



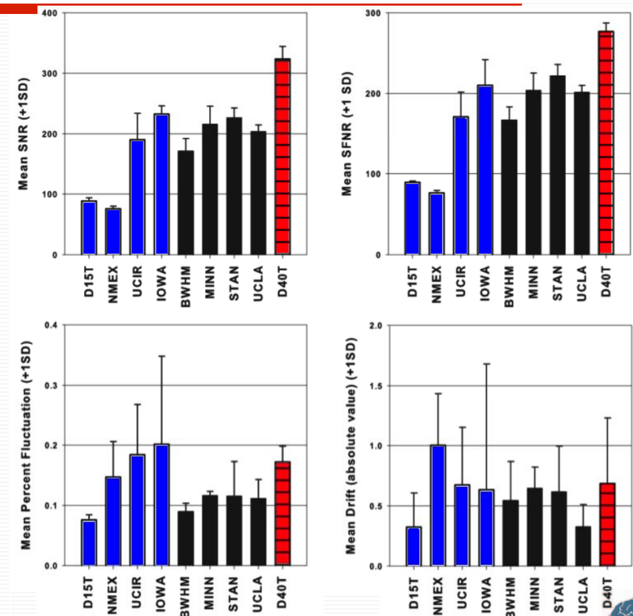


ROI Length

ROI Length

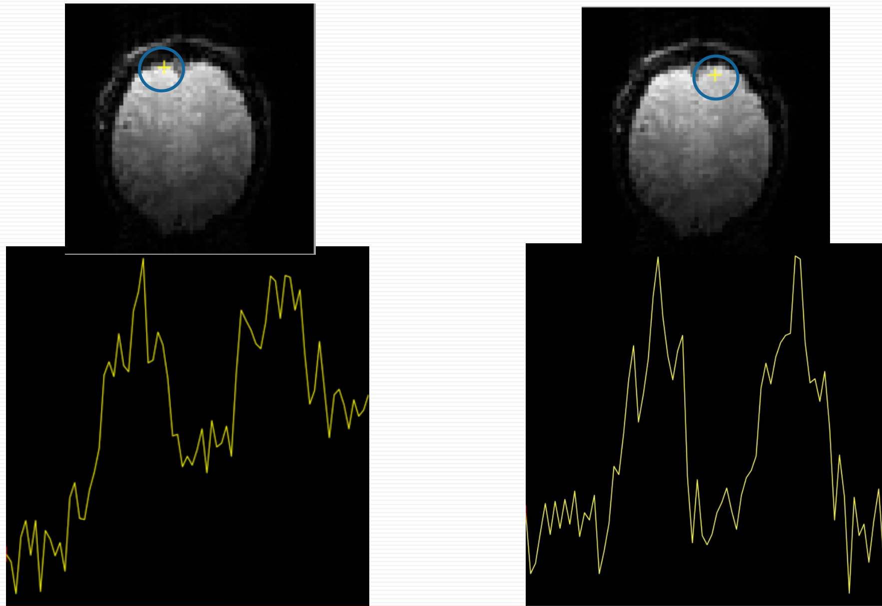
Friedman and Glover, JMRI 23:827

Scanner Comparisons



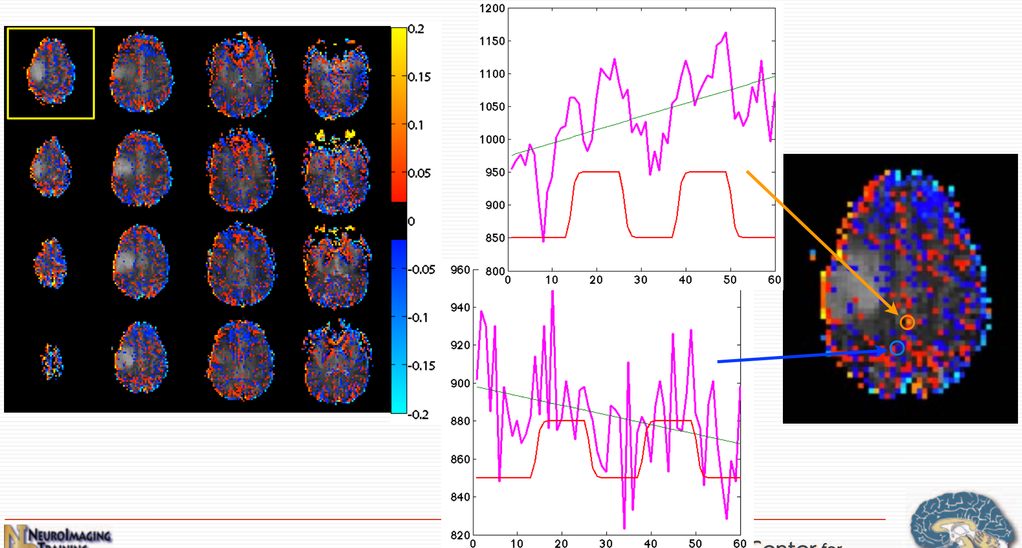
Friedman and Glover, JMRI 23:827

“Drift”

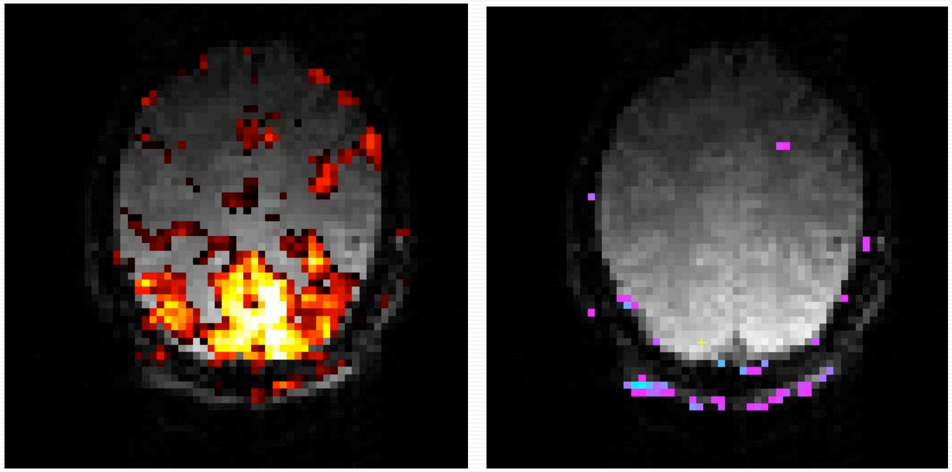


Instrument Variation

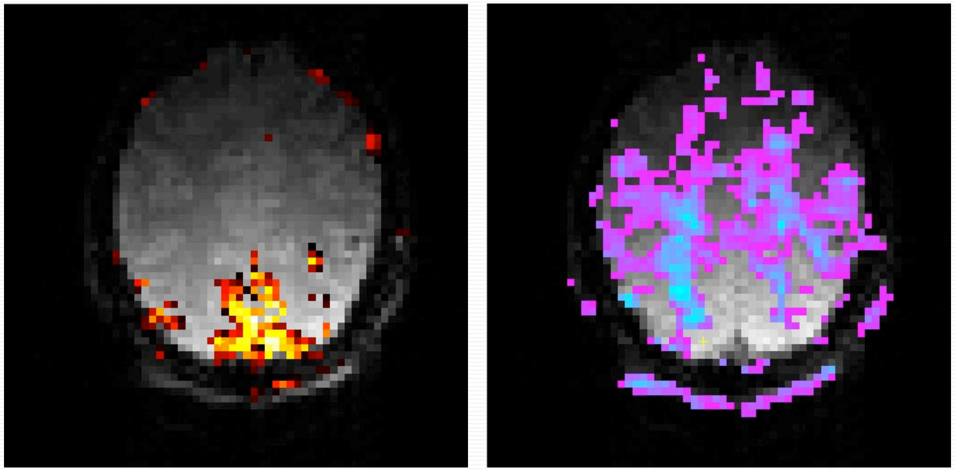
- 2. The mystery of scanner drift.



Global Mean Scaling - OFF



Global Mean Scaling - ON



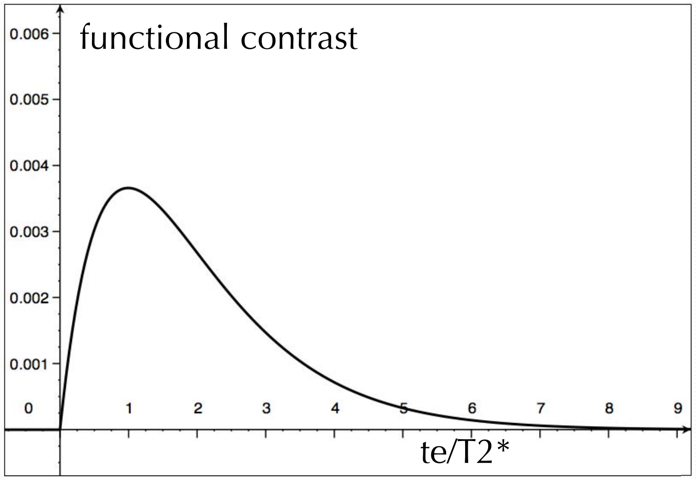
Thermal Noise

- 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

NeuroImaging Training Program

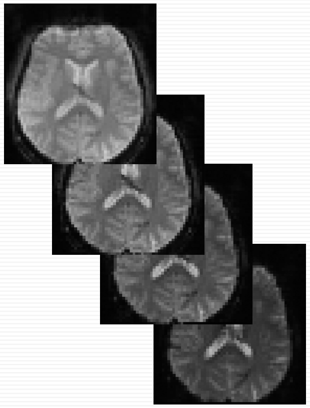
Parameter Optimization

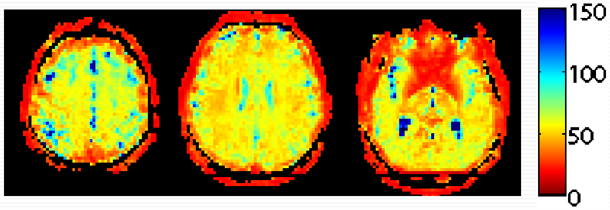
- Best BOLD contrast when te=T2*



Parameter Optimization

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

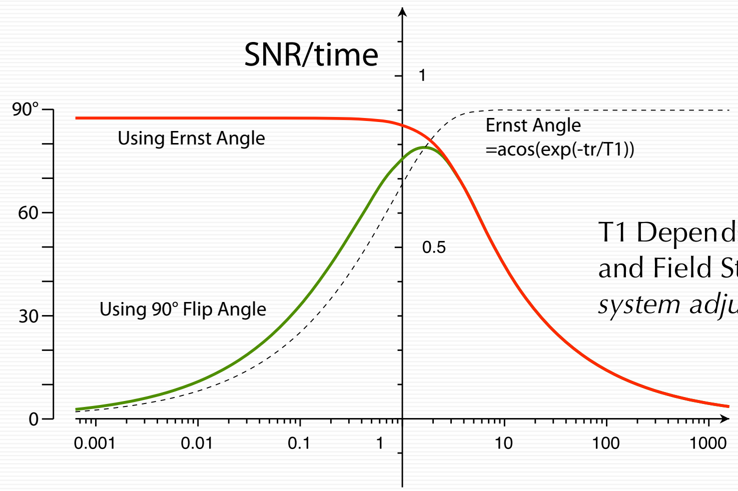




T2* is both an *Instrument* Parameter and a *Physiological* Parameter

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*)

Exercise in Afternoon Lab

Other Parameters…

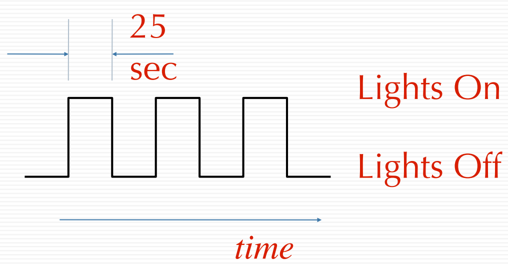
- Voxels Should be Large enough that:
  - Thermal Noise << Physiological Fluctuations
- For Best Signal with Arbitrary Slice Orientation:
  - Voxels Should be Isotropic
- With Gradient Echo Scans (most BOLD):
  - Signal Falls (much) more than linearly with Slice Thickness
- Do not Confuse Signal to Noise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

Some Theoretical Considerations

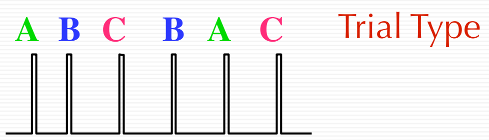
- Study Designs:
  - *Blocked*
  - *Single Trial*
- Predicting Responses
- Sources of Variance
- Resolution Limits:
  - *Temporal*
  - *Spatial*

Blocked vs. Single Trial

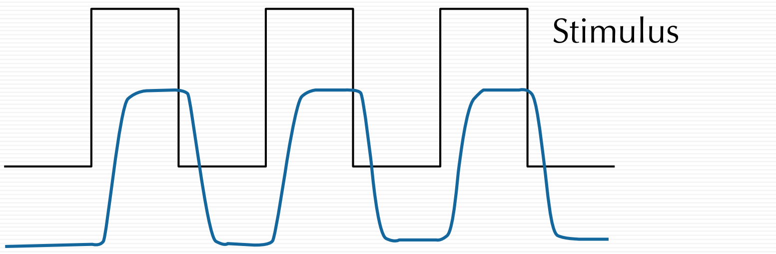
Typical Blocked Design



Typical Single Trial Design

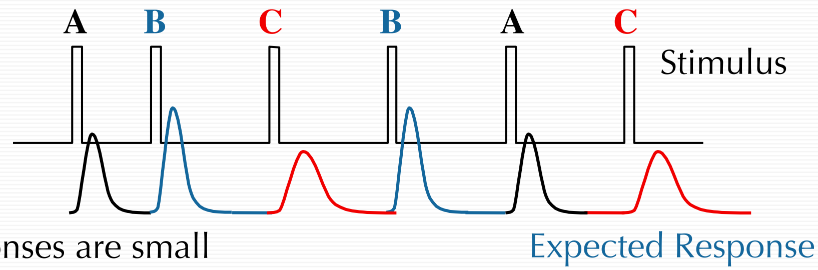


Blocked Experiments



- •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

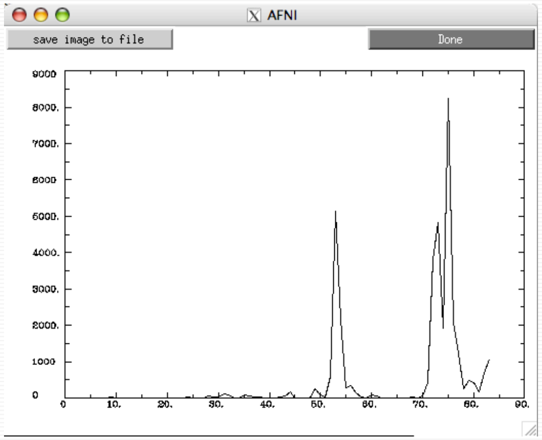
Single Trial Designs



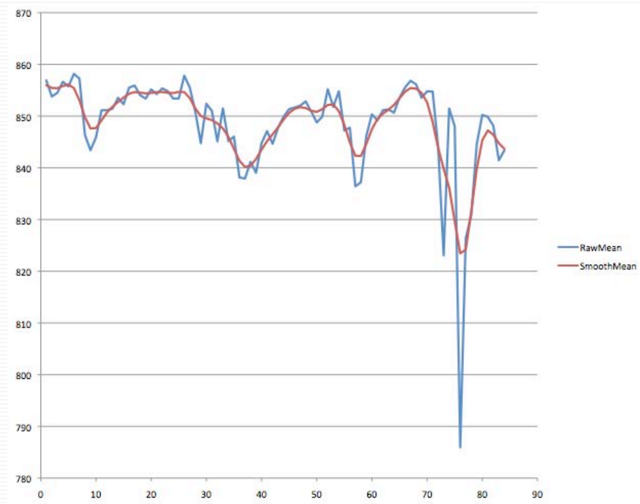
- Responses are small
- Useful contrast/noise is low
- Data are more Challenging to Analyze
- Exact Time course is Modeled or a Dependent Variable
- Suitable for Randomized Stimulus Designs

AFNI 3dToutcount

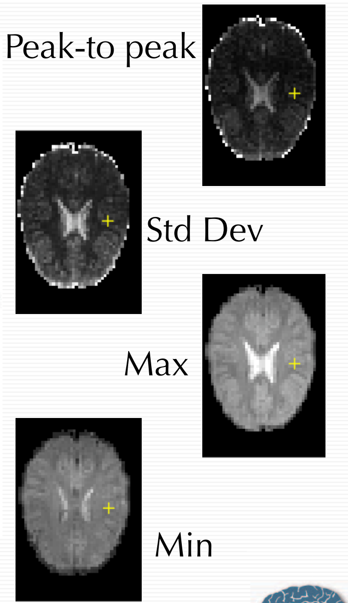
```
3dToutcount -automask KidAsImg.hdr | 1dplot -stdin
```



P2P



870
860
850
840
830
820
810
800
790
780
0
10
20
30
40
50
60
70
80
90
RawMean
SmoothMean



Peak-to peak
Std Dev
Max
Min

What really happened?