G16.4426/EL5823/BE6203 Medical Imaging

Image Quality, Diffusion MRI, Functional MRI

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Outline

- Review of Image Reconstruction
- Image Quality
  - Image Resolution
  - Common Artifacts
  - Signal-to-Noise Ratio (SNR)
- Diffusion MRI
- Functional MRI

How is an image produced?

Earlier we discussed the origin of the MR signal and how it may be manipulated to produce different types of signal contrast

We saw that the origin of the MR signal involves:

- Polarization of spins by a static B_0 field in the z direction
- Excitation of spins by a rotating B_1 field in the x-y plane
- Detection of the emitted signal by a receiver coil

We also saw that the emitted signal could be sensitized to tissue-dependent properties such as relaxation times to achieve signal contrast among different tissues and lesions

To produce an image, however, we need to know **where** the signal originates, and know it **with high resolution**

This is not possible using just the main magnet and the RF excitation and receiver coils, however, since they encompass the entire body (or body part) of interest

Imaging: the solution

A solution to the problem of mapping the spatial distribution of the MR signal was invented by Paul Lauterbur and Peter Mansfield, for which they won the Nobel Prize in 2003.

They observed that the frequency of precession is a very precise measure of the ***local magnetic field*** at the site of the spins

Therefore, by introducing magnetic field gradients, the frequency could be used to identify the ***position*** of the spins

Magnetic field gradients alter the precession frequency of the spins in a spatially-dependent manner

They are used in two different ways to produce an image:

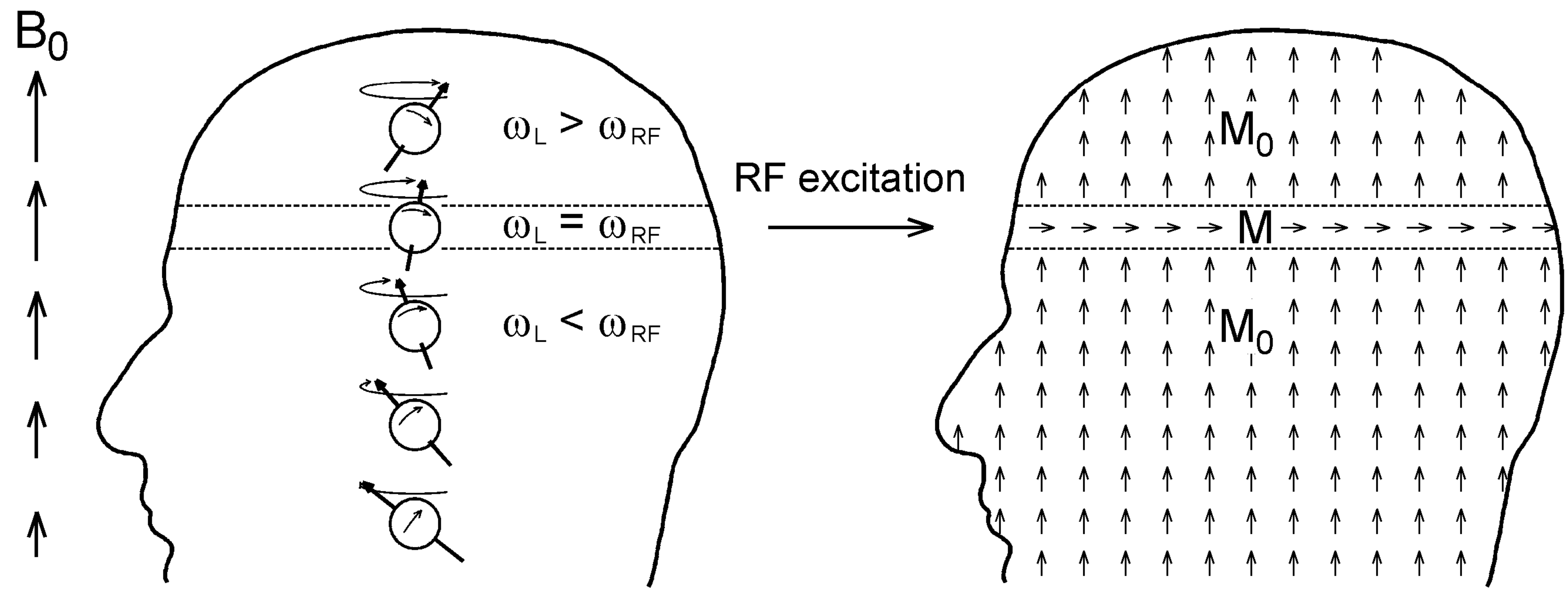
- Selective excitation

  When applied ***during*** excitation, magnetic field gradients ensure that only certain spins are excited

- Spatial encoding

  When applied ***after*** excitation, magnetic field gradients can be used to encode spatial information in the signal via the spins' frequency and phase

Slice-selective excitation



A magnetic field gradient is applied during the RF excitation pulse

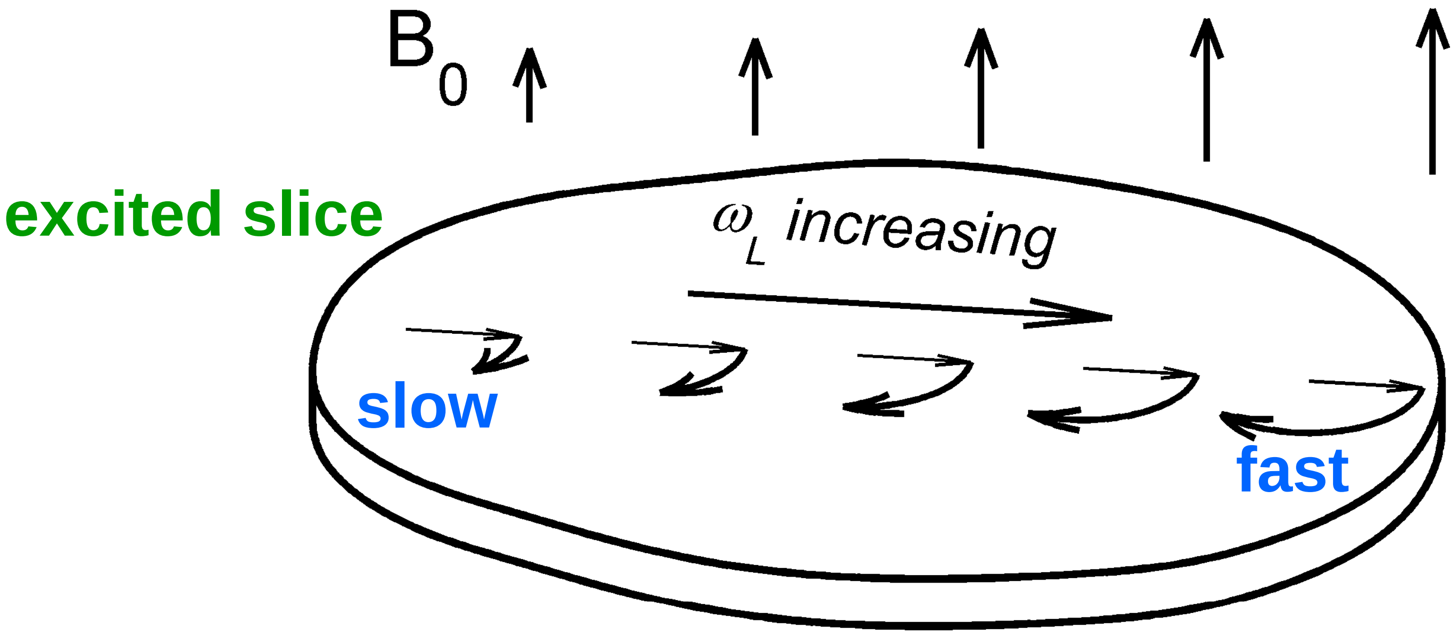
(Note that in the above diagram the gradient is applied in the same direction as B_0, but it can in practice be along any direction)

The gradient alters the Larmor frequency ω_L of the spins along the direction of the gradient

Only those spins whose Larmor frequency equals the frequency of the RF pulse $\omega_L = \omega_{RF}$ will be excited

Such spins lie in a ‘slice’ of tissue perpendicular to the gradient

Frequency encoding

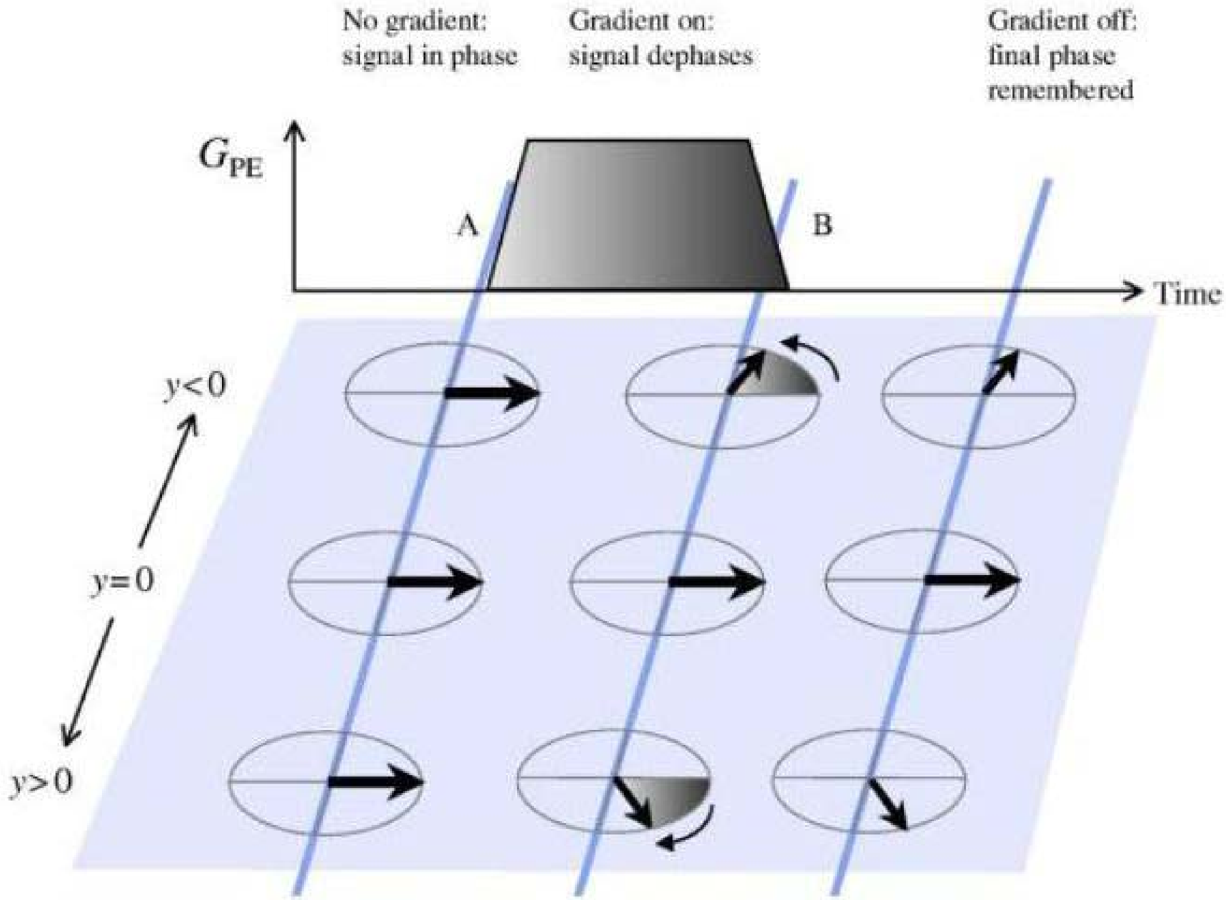


A magnetic field gradient is applied during the data acquisition

The gradient alters the Larmor frequency ω_L of the spins in a spatially-dependent manner

The frequency of the signal emitted by each spin will therefore depend on its location along the direction of the gradient

The frequency thus provides a 'label' to identify the spins' location

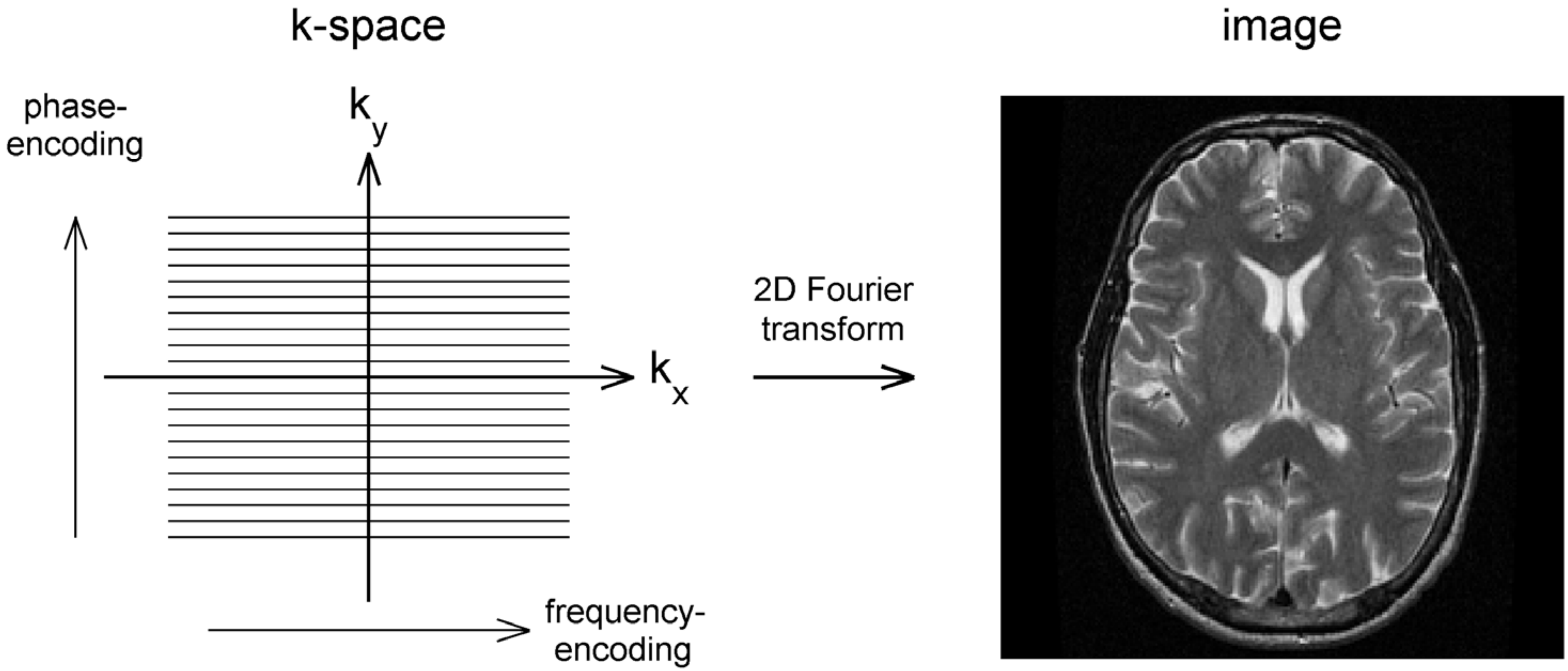


A magnetic field gradient is applied in the remaining direction for a short period ***after*** excitation but ***before*** data acquisition

The gradient imparts a spatially-varying phase shift to the spins

During the subsequent data acquisition period, the spins along any line in the phase-encoding direction will precess with identical frequencies but different phases

Image reconstruction



In this lecture we will discuss in more detail how the sampling of k-space affects the field of view and resolution of the image

Spatial frequency components of an image

Low spatial frequency components capture the overall signal intensity and shading. Higher spatial frequency components describe the fine structure and edges of an object.

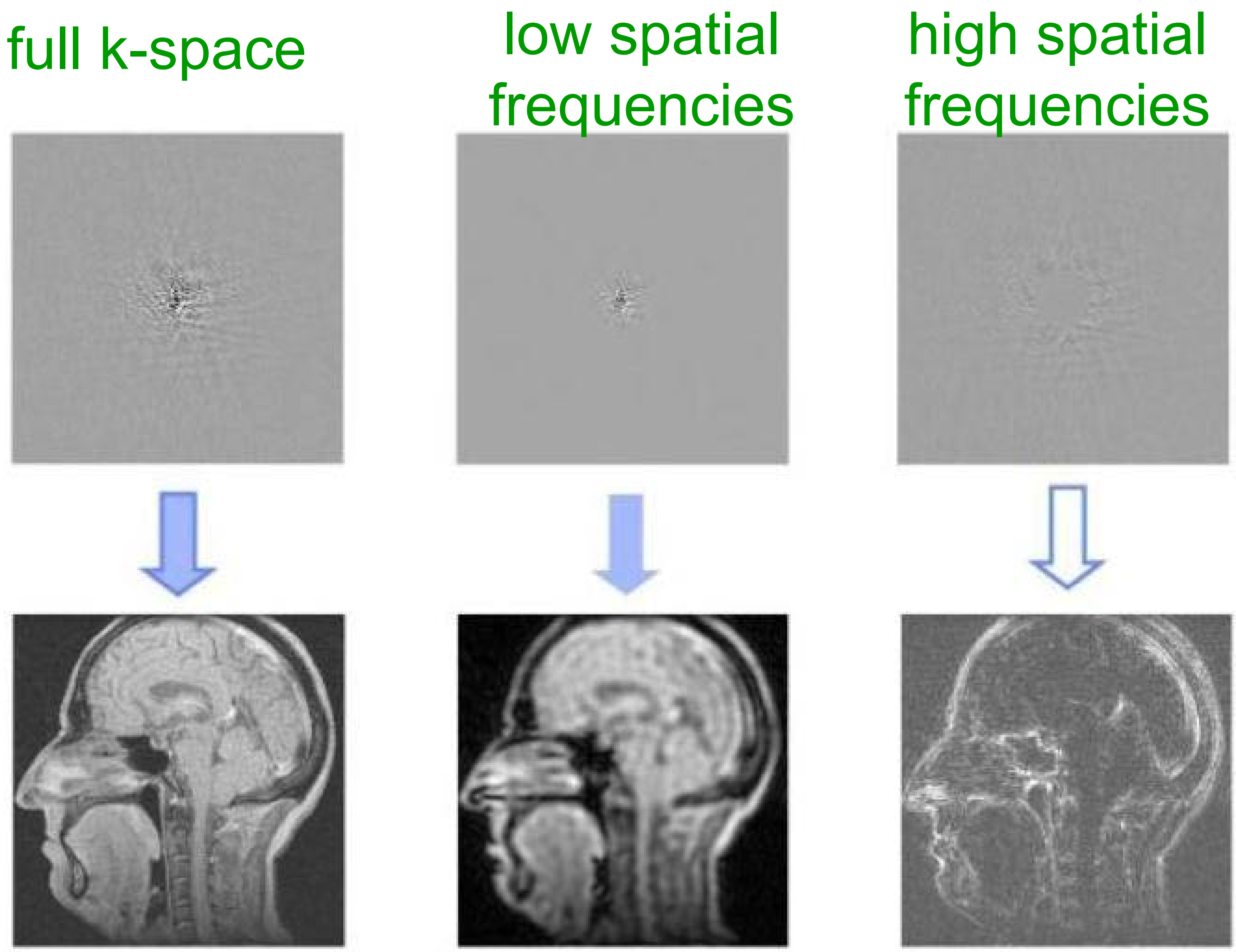


Image ≠ signal distribution

There are however other important differences between the image we see on the MRI console and the actual signal distribution. Some arise from the fact that we can never sample all of k-space.

The pair of equations relating the k-space data $s(k_x,k_y)$ and the signal distribution are both ***continuous*** Fourier transforms

$$s(k_x, k_y) = \int S(x, y)\exp\left[-i2\pi(k_x x + k_y y)\right]\mathrm{d}x\mathrm{d}y$$

$$S(x, y) = \int s(k_x, k_y)\exp\left[i2\pi(k_x x + k_y y)\right]\mathrm{d}k_x\mathrm{d}k_y$$

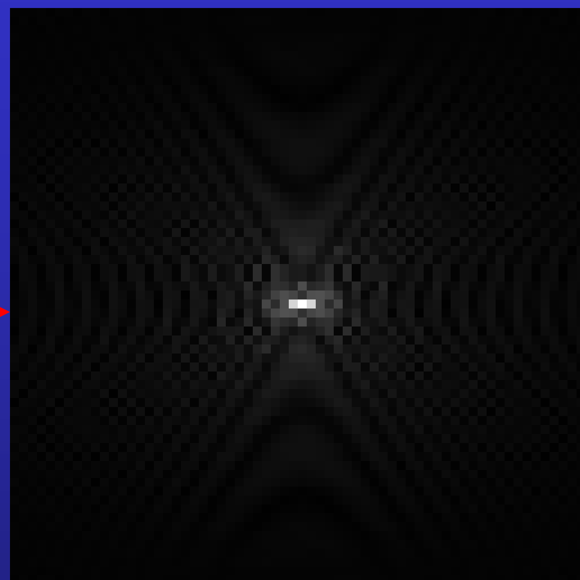
i.e. they involve integrals, not sums, and the ranges are infinite.

In practice, however, k-space is sampled discretely, so the second equation is approximated by a discrete Fourier transform.

Image reconstruction

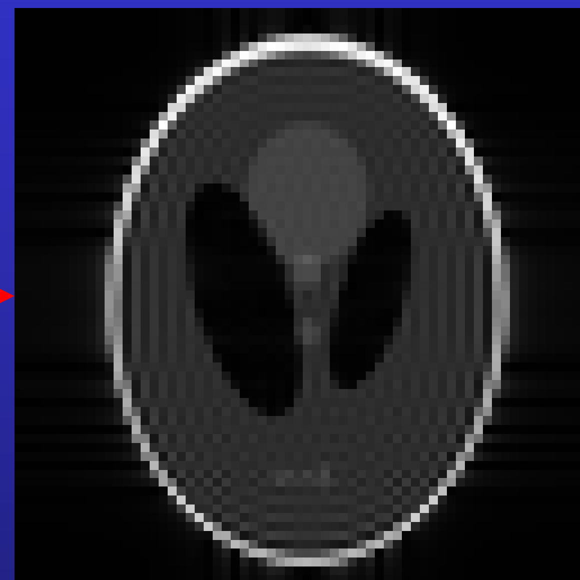
Phantom
in physical space

Raw data
in k-space



64x64

reconstructed
image



64x64

The fact that k-space is sampled discretely has important implications for the image that we finally obtain. It affects the resolution and field of view, and it can also cause image artifacts

Mathematical description of image reconstruction

For simplicity we shall treat the mathematical description of image reconstruction in one dimension. The extension to two dimensions is trivial, but unnecessarily complicates the equations.

K-space is usually sampled more coarsely in the phase-encoding direction, so the implications of discrete sampling are more evident in that direction. We will therefore consider the y-direction.

In 1D the pair of equations relating the k-space data and the signal distribution along the y direction are

$$s(k_y) = \int S(y)\exp\left[-i2\pi k_y y\right]\mathrm{d}y$$

$$S(y) = \int s(k_y)\exp\left[i2\pi k_y y\right]\mathrm{d}k_y$$

The image is reconstructed from the discretely sampled data

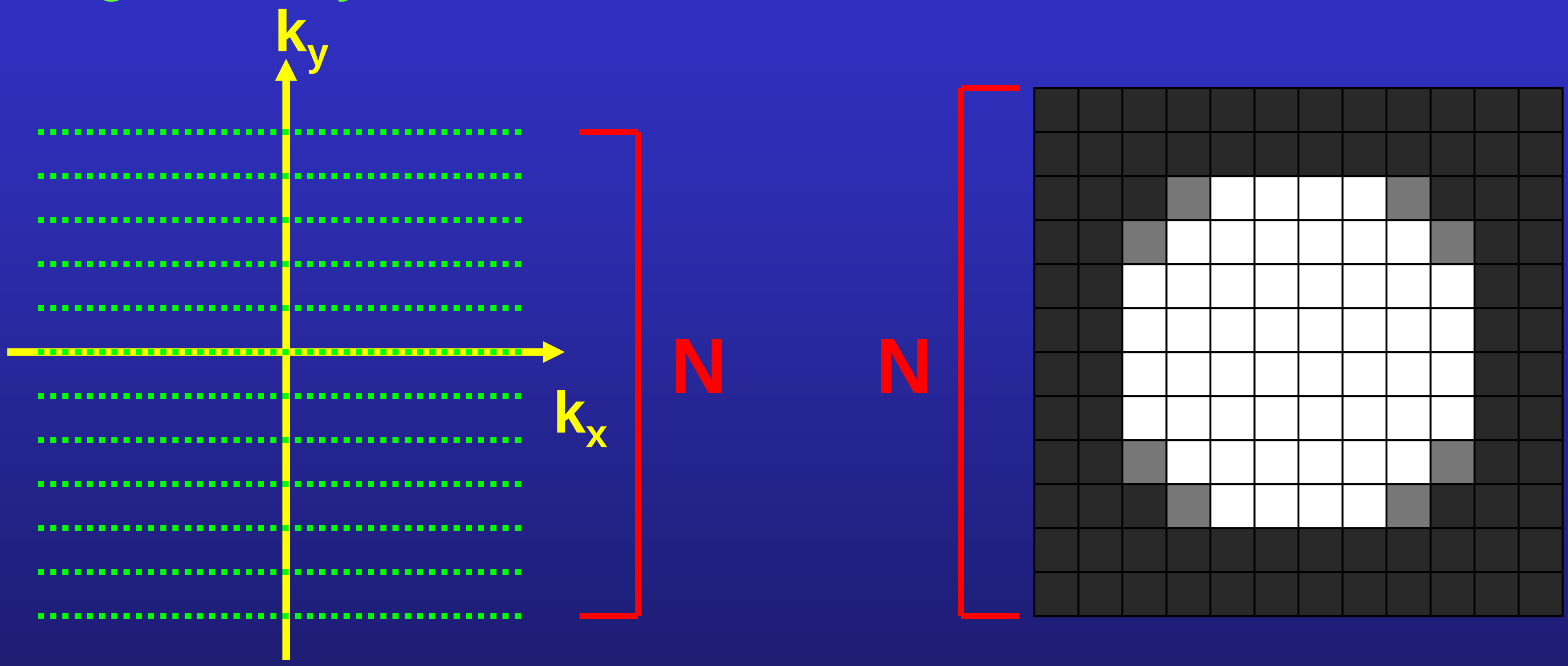
$$\hat{S}(m\Delta y) = \frac{1}{N}\sum_{n=-N/2}^{N/2-1} s(n\Delta k_y)\exp\left[i2\pi n\Delta k_y m\Delta y\right]$$

where –N/2 < m < N/2 -1

Matrix size

The first observation about k-space sampling is that

The number of k-space lines N determines the number of pixels in the image in the y-direction.

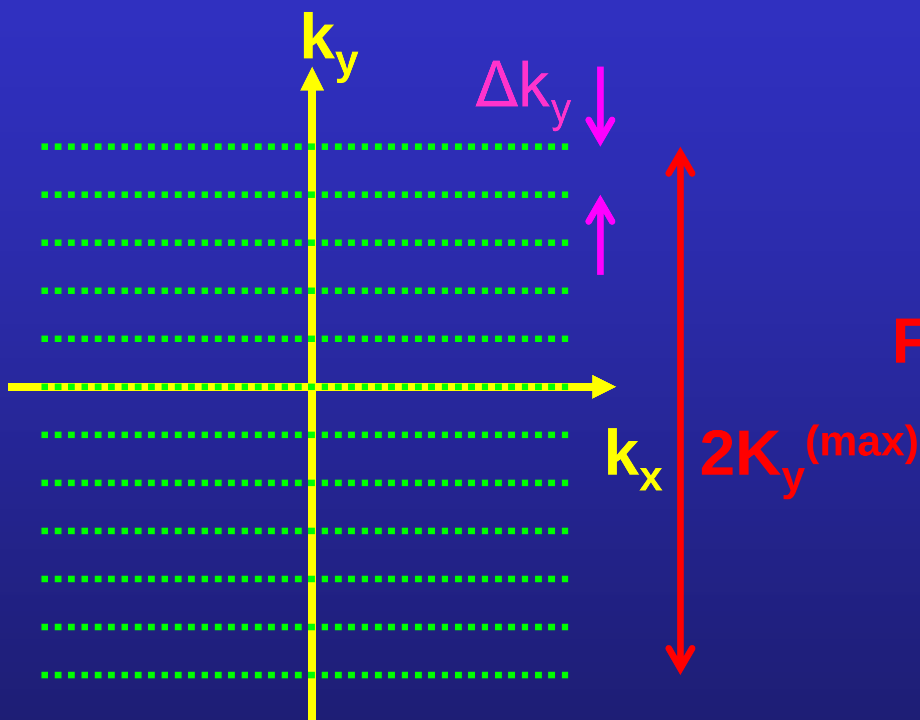


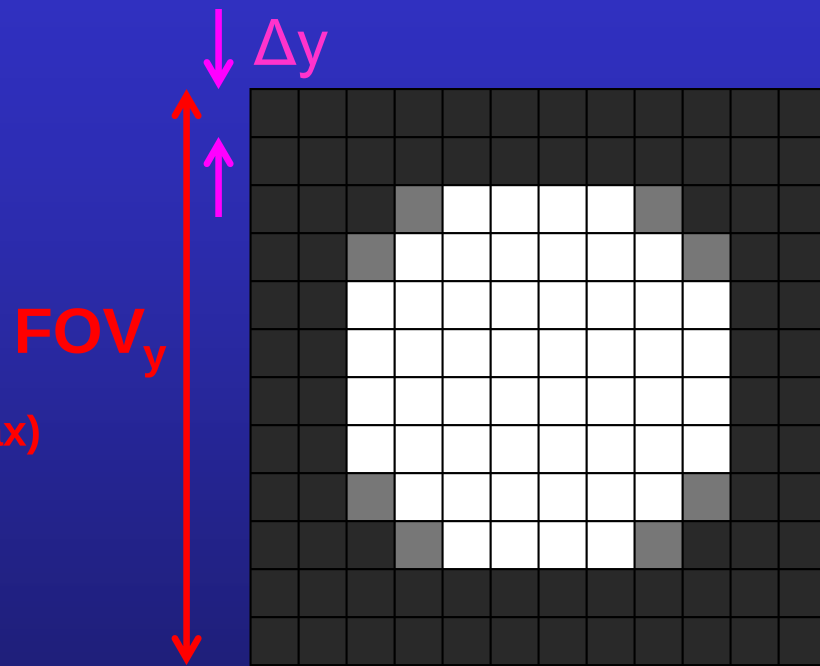
More generally, in 2D:

The matrix size of k-space = the matrix size of the image

Definitions

To express the relationships between the k-space and image-space parameters, we first need to establish some definitions:



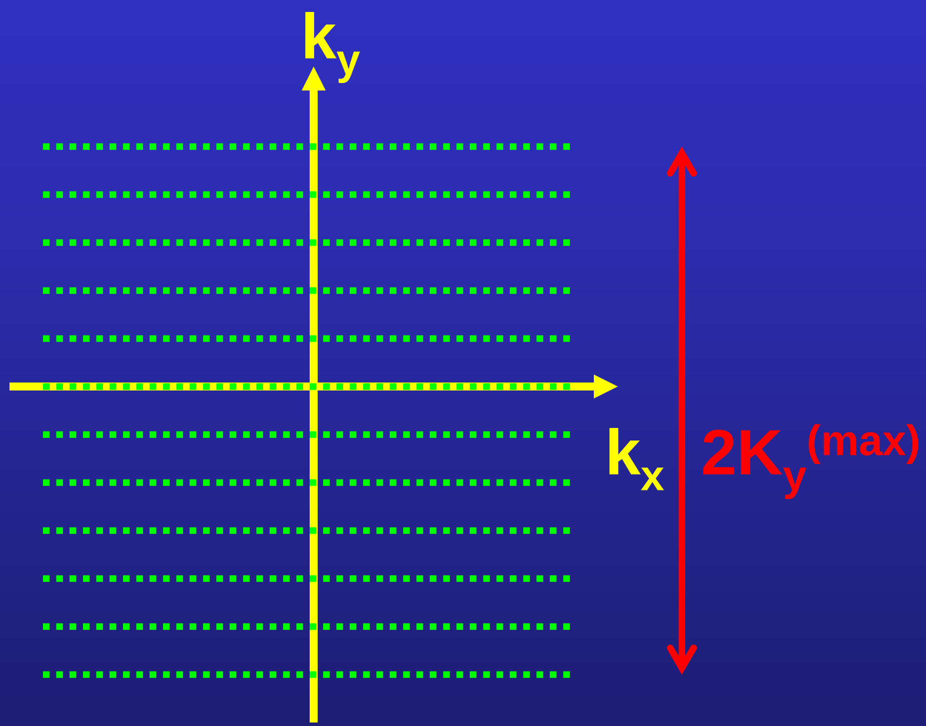


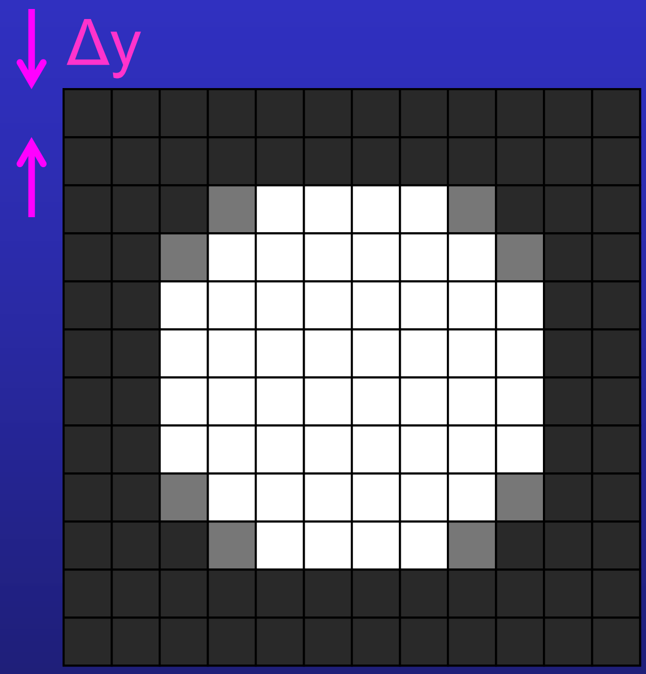
$\Delta k_y = 2K_y^{(max)} / N$

$\Delta y = FOV_y / N$

Relationship between spatial resolution and K_{max}

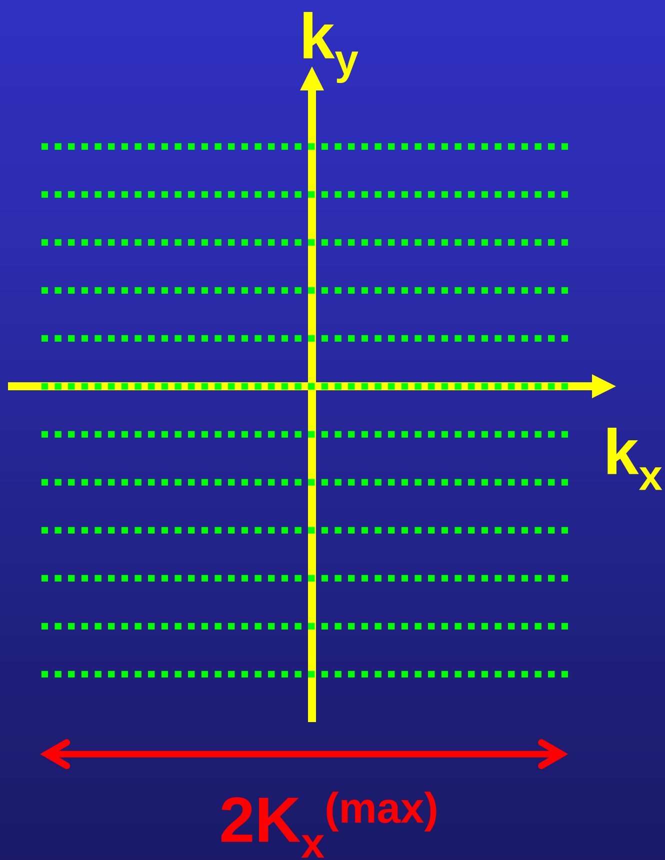
K_{max} signifies the highest spatial frequency we sample. If we sample out to higher spatial frequencies, we get better resolution in the image (i.e. smaller Δy).

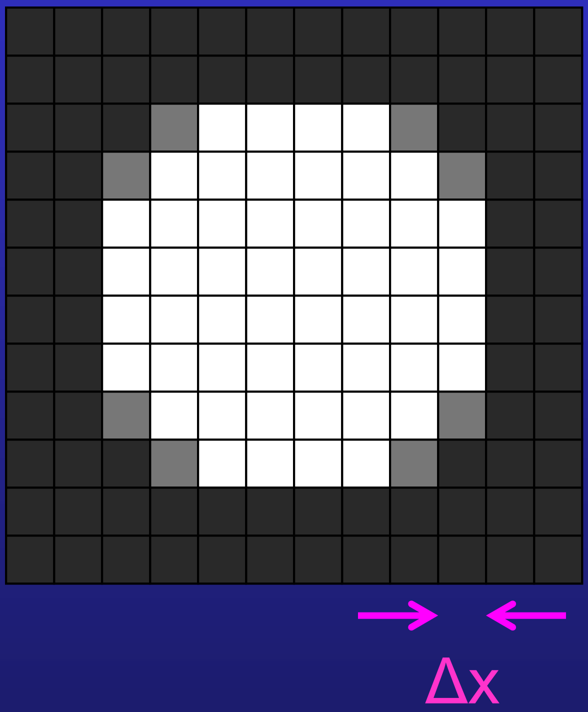




$$\Delta y = \frac{1}{2K_y^{(max)}}$$

…and similarly in the x direction

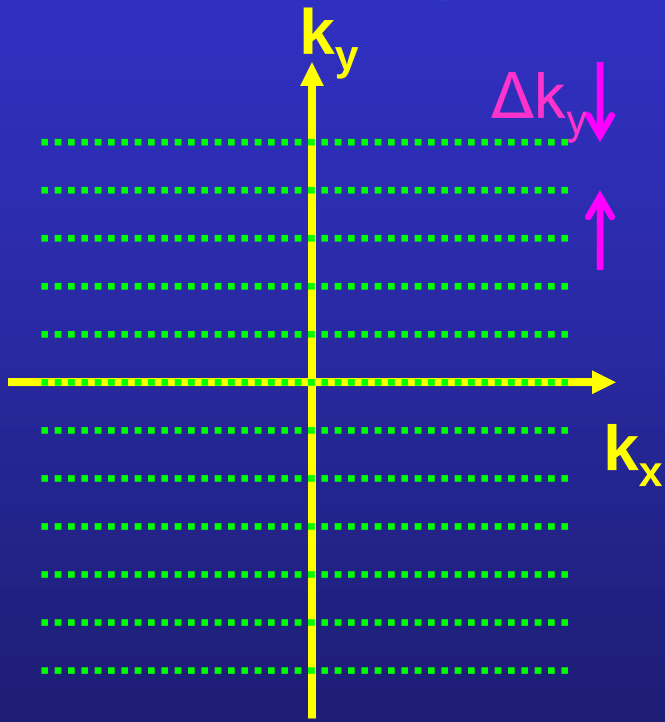


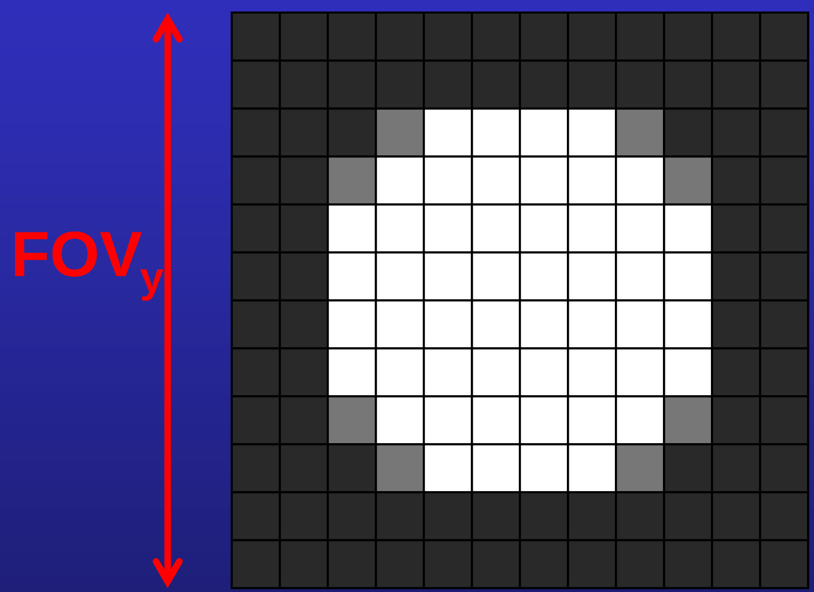


$$\Delta x = \frac{1}{2K_x^{(max)}}$$

Relationship between k-space sampling density and FOV

This is analogous to the previous relationship. To cover a larger FOV, we require higher sampling density (smaller Δk_y).

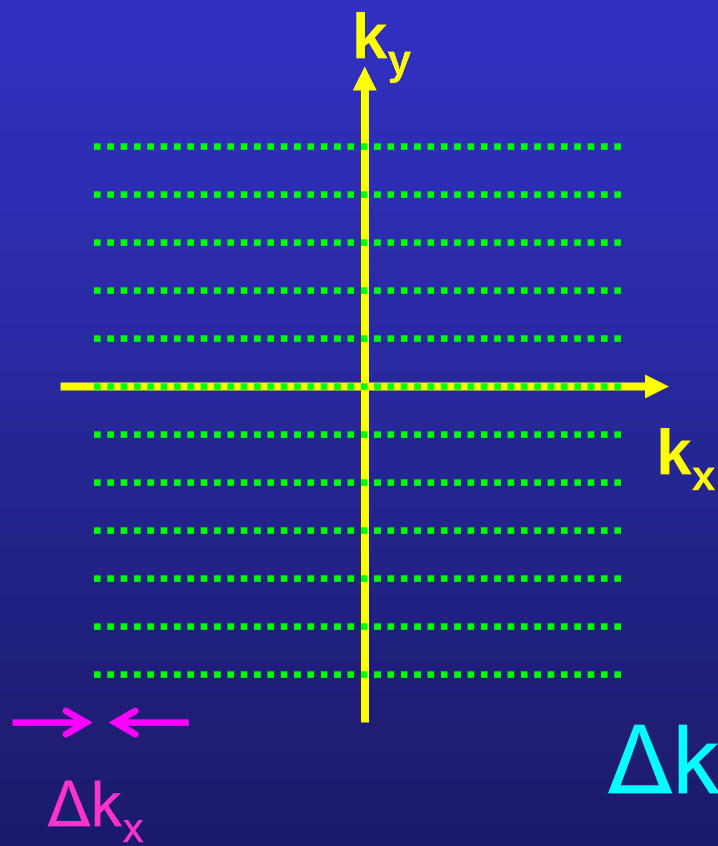




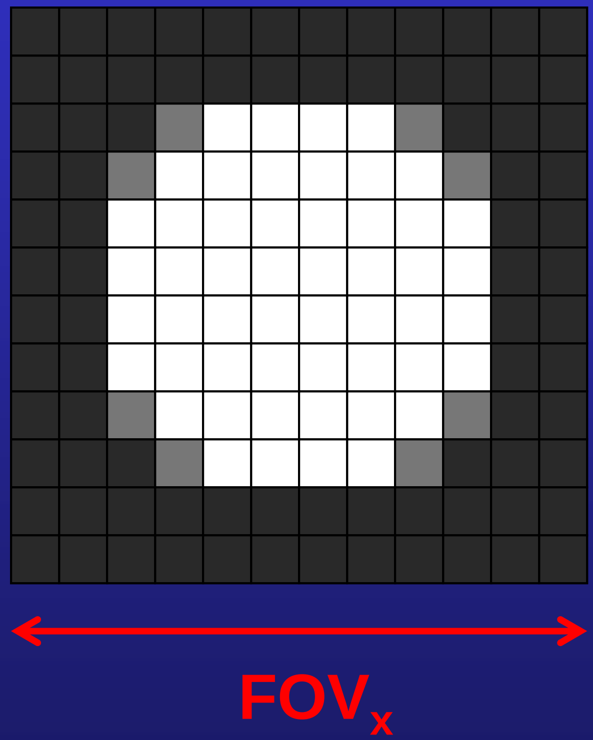
$$\Delta k_y = \frac{1}{FOV_y}$$

…but in the x direction k-space is oversampled

In the x direction, it does not cost us anything to use a higher sampling density – we simply sample the signal at a higher rate



$$\Delta k_x < \frac{1}{FOV_x}$$



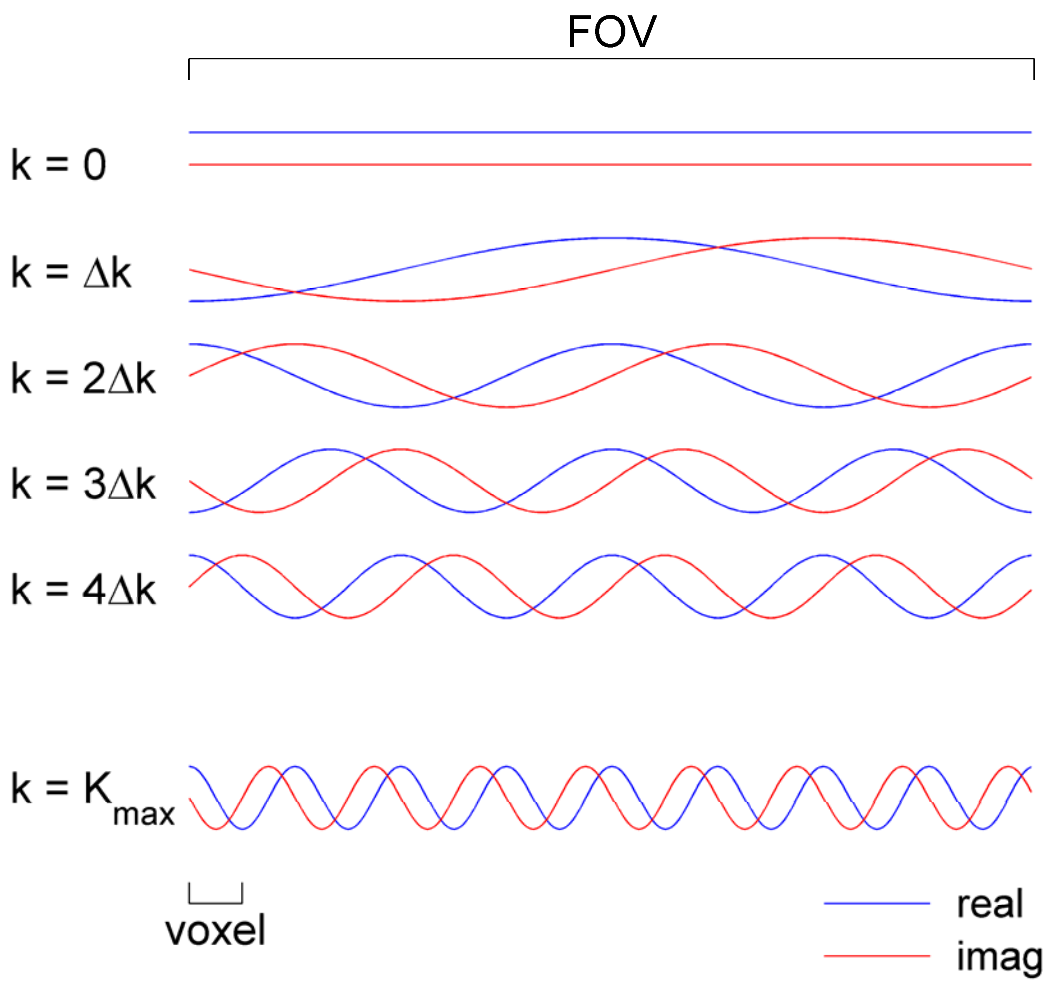
Typically k-space is oversampled in the x direction

We will see why this is useful when we consider aliasing

Observation

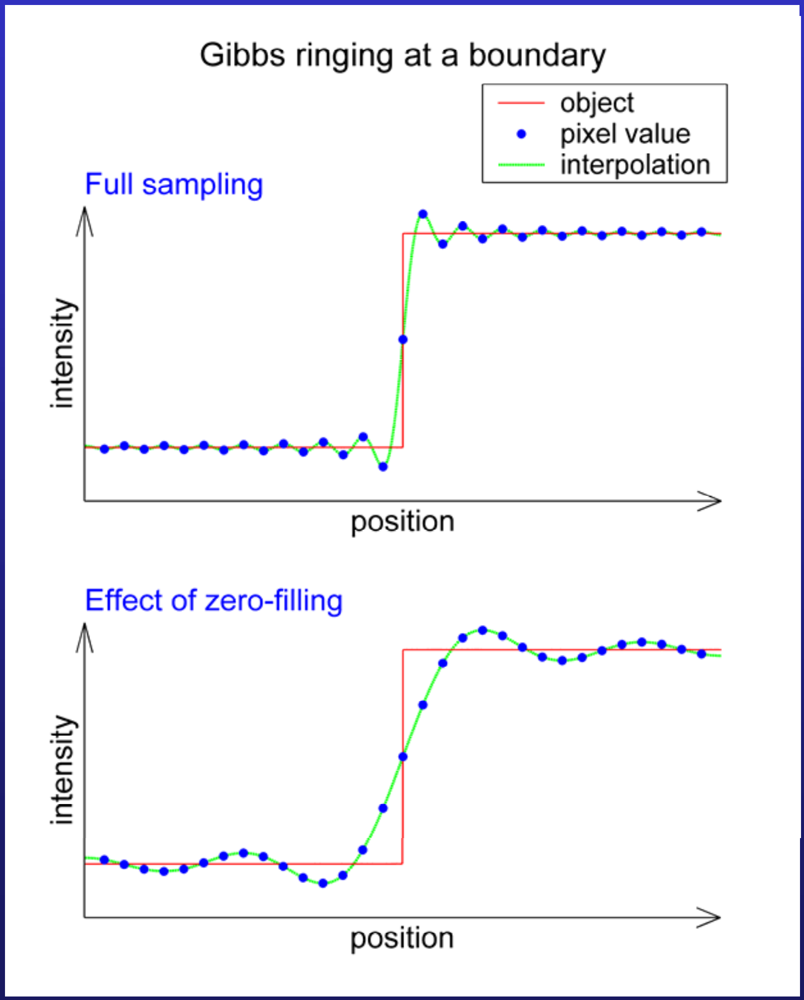
Each additional step Δk introduces an additional phase change of one full cycle (2π) across the entire FOV.

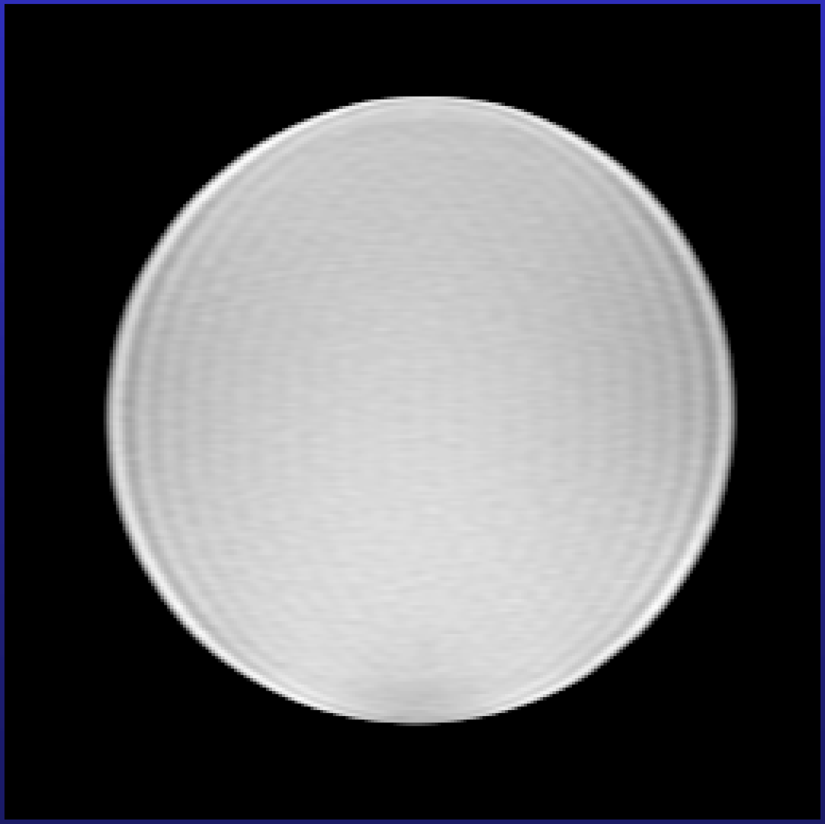
Between $-K_{max}$ and K_{max} a phase change of one full cycle (2π) is introduced across each voxel



Truncation artifacts (Gibbs ringing)

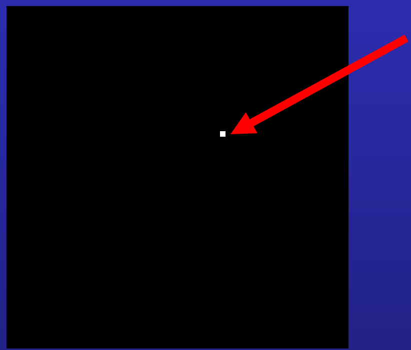
Truncation artifacts occur when there is fine structure in the object (i.e. high spatial frequency components) that the k-space sampling does not capture because K_{max} is not large enough.

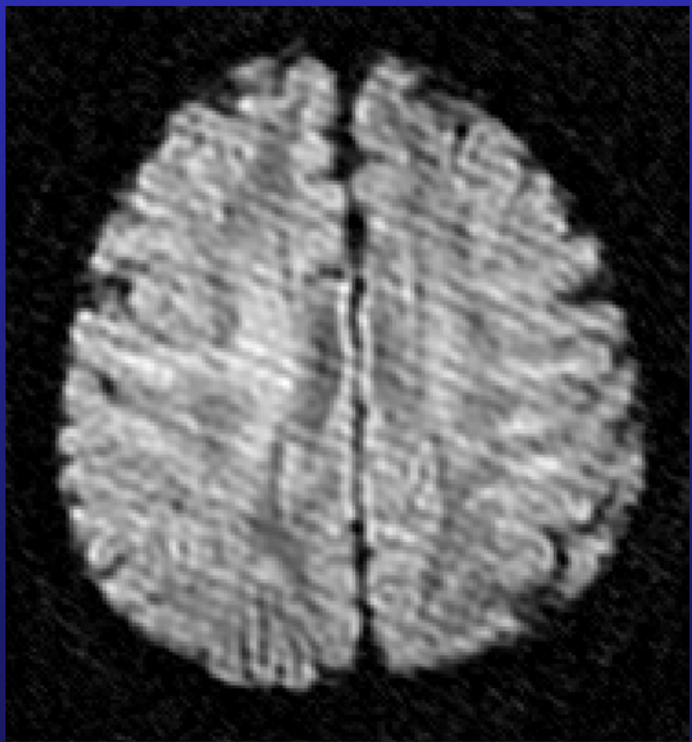




Spike artifacts

Spikes arise when one or more individual k-space components is corrupted. They usually occur due to spurious electrical discharges or arcing, e.g. electrical cables rubbing against each other, or burnt-out light bulbs in the scanner suite





Aliasing (wrap-around)

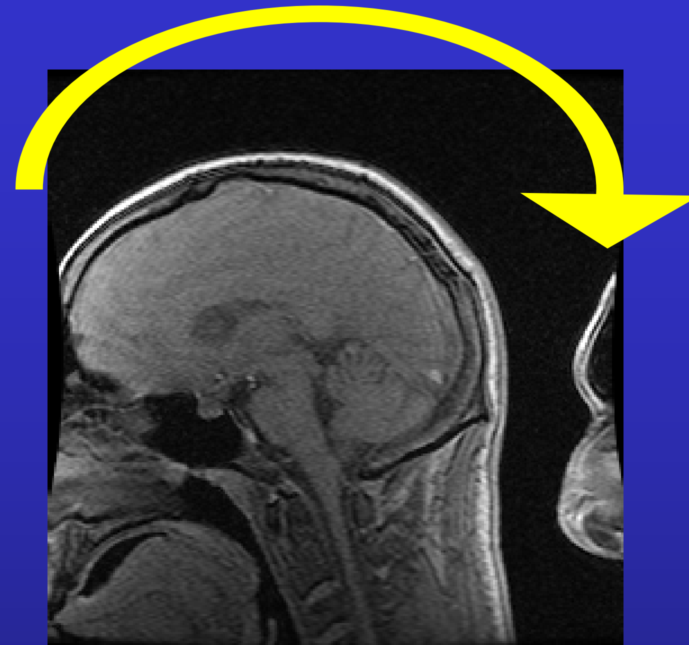
If there is tissue outside the prescribed field of view in the phase-encoding direction then tissue that lies outside the FOV on one side is 'wrapped' into the FOV on the other side

This occurs because of the finite sampling density (i.e. the discrete step size Δk_y)

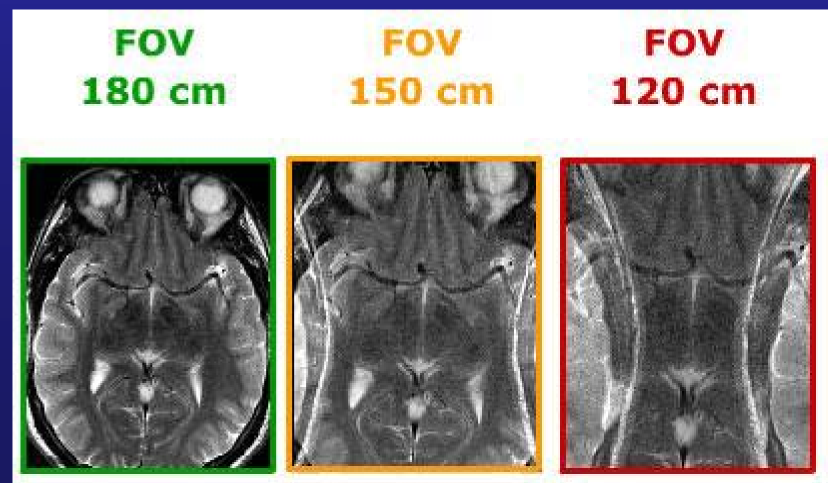
Δk_y is related to the FOV through:

$$\Delta k_y = \frac{1}{FOV_y}$$

so taking finer steps Δk_y is equivalent to using a larger FOV



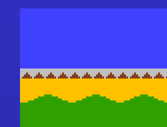
off-center FOV



FOV too small

Cause of aliasing

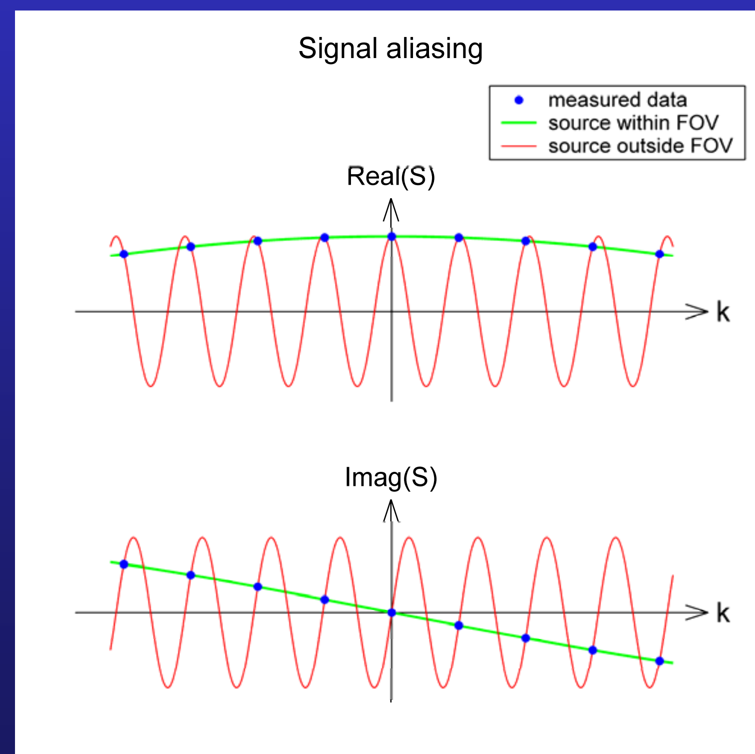
Aliasing is analogous to the effect that you see in old Western movies when the frame rate is not high enough to capture the motion of the wagon wheels; when the wheels reach a certain speed they appear to move backwards.



In phase-encoding, the position of the spins is determined by the ***phase change*** between successive acquisitions.

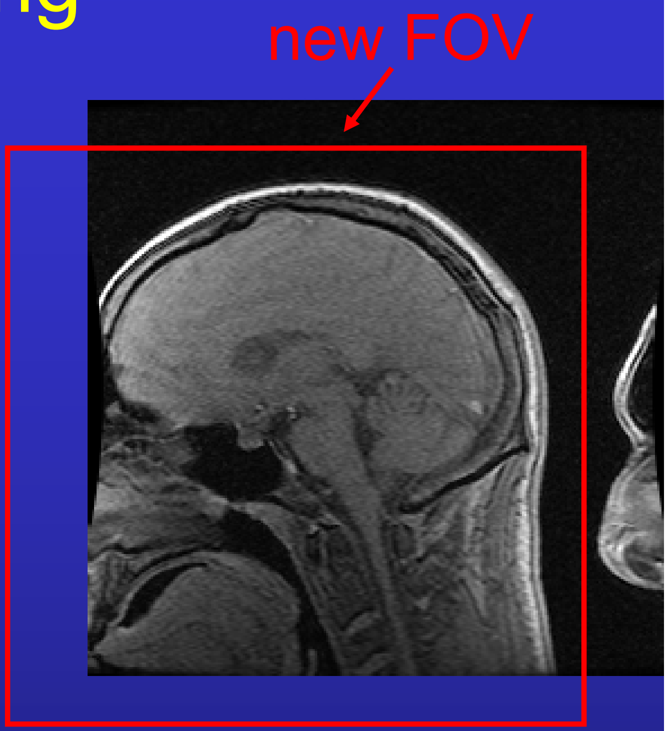
A phase change of $\Delta\varphi = 2\pi + \theta$ (from tissue outside the FOV) cannot be distinguished from a phase change of $\Delta\varphi = \theta$ (inside the FOV).

The reconstruction algorithm therefore attributes the signal to the point inside the FOV.



How to avoid aliasing

Aliasing can be avoided by centering the FOV on the sample and ensuring that it is large enough to encompass all the tissue.



Aliasing does not occur in the frequency encoding direction since the raw data are band-pass filtered with an anti-aliasing filter and then oversampled in that direction, i.e. collected with a smaller step size

$$\Delta k_x < \frac{1}{FOV_x}$$

Note that oversampling in the readout direction does not increase the scan time. The data simply have to be sampled at a higher rate. Typically the readout data are oversampled by a factor of 2.

Oversampling can be used in the phase-encoding direction but it increases scan time, since more lines of k-space need to be acquired

Signal, Noise and SNR

- Signal (S) is the part of the measurement (I) we wish to study
- Noise (N) is the part of the measurement we do not wish to study

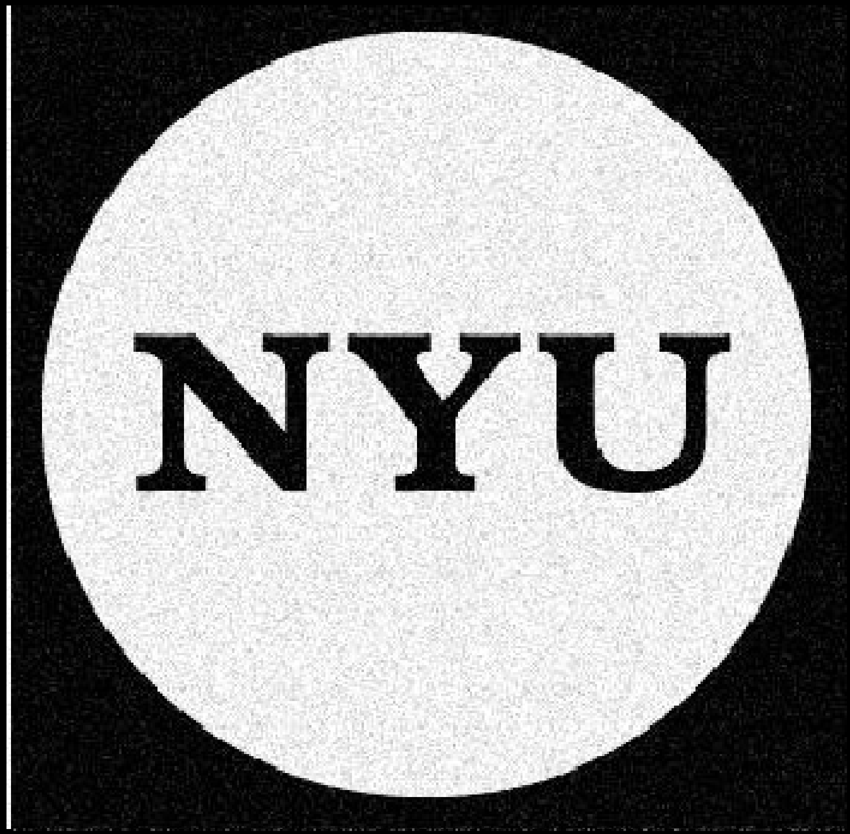
$$I = S + N$$

- Signal-to-noise ratio (SNR) is a measure of the relative contribution of each part to the measurement

$$SNR = \frac{\text{mean}(S)}{\text{stdev}(N)}$$

SNR and Visual Interpretability

- SNR is a criterion for image quality



SNR = 256

The NMR Signal

- The available signal depends on the difference between the spins in the up and low energy states (i.e. the net magnetization vector)
  - Varies for different MR nuclei (^{1}H, ^{23}Na, ^{13}C,…)
  - Increases with abundance of the nuclei in tissues
  - Increases at larger magnetic field strength

Hydrogen (proton imaging)	0.5 T	1.5 T	3.0 T
	1.65 ppm	4.94 ppm	9.88 ppm

Noise Sources

- Johnson noise due to thermal agitation of electrons in coil conductors and sample (per unit bandwidth):

$$\sqrt{4k_B TR}$$

Boltzmann constant

Temperature of the object

Noise equivalent resistance

- The noise equivalent resistance models conductor losses, radiation losses, sample losses
- *NF* is the noise figure of the system and models the noise introduced by the receive chain (preamplifiers, cables, circuitry, etc.)

Noise in the Image

- Looking only at how the noise component is transformed:

$$\mathrm{E}(p\Delta x, q\Delta y) = \frac{1}{N_x N_y} \sum_{p'=-N_x/2}^{N_x/2-1} \sum_{q'=-N_y/2}^{N_y/2-1} \varepsilon(p'\Delta k_x, q'\Delta k_y) e^{i2\pi\left(\frac{pp'}{N_x}+\frac{qq'}{N_y}\right)}$$

- Taking the variance of both sides yields:

$$\sigma_0^2 = \frac{\sigma_k^2}{N_x N_y}$$

→ Noise variance at any voxel in the image is $N_x N_y$ times smaller than in the raw data (detected signal) and is the same for all voxels

- SNR is calculated using noise standard deviation $\sqrt{\sigma_0^2}$, so it is proportional to the square root of the number of acquired samples
  - SNR increases with matrix size if FOV is increased (i.e. same voxel size)
  - Using iPAT (acceleration factor) = *R* reduces the acquired data points in the phase encoding direction by N_y / R, increasing the noise standard deviation by $\sqrt{R}$

→ $$SNR_{iPAT=R} = \frac{SNR}{\sqrt{R}}$$

Increasing SNR by Averaging

- The MRI system allows repeating the entire image experiment (averaging) to improve SNR
- The mean of the averaged signal (raw data) is the same:

$$s_{m,ave}(k) = \frac{1}{N_{ave}} \sum_{i=1}^{N_{ave}} s_{m,i}(k) = \frac{1}{N_{ave}} N_{ave} s(k) = s(k)$$

- The variance of the averaged signal is smaller by N_{ave}:

$$\sigma_{m,ave}^2 \equiv \mathrm{var}\left(s_{m,ave}(k)\right) = \frac{1}{N_{ave}^2} \sum_{i=1}^{N_{ave}} \mathrm{var}\left(s_{m,i}(k)\right) = \frac{\sigma_m^2(k)}{N_{ave}}$$

Assuming the noise is statistically independent between acquisitions

- The SNR of the k-space signal is larger by the square root of N_{ave}:

$$SNR(k) = \frac{s_{m,ave}(k)}{\sqrt{\sigma_{m,ave}^2(k)}} = \sqrt{N_{ave}} \frac{s(k)}{\sigma_m(k)}$$

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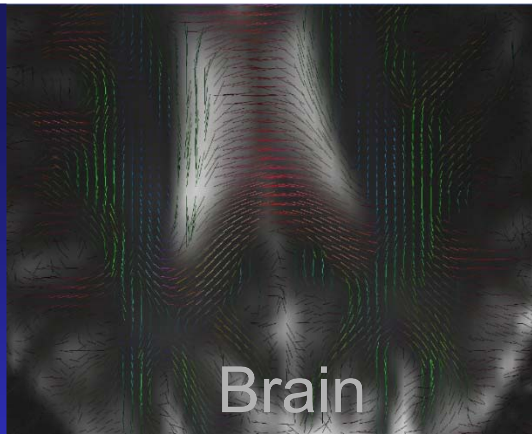
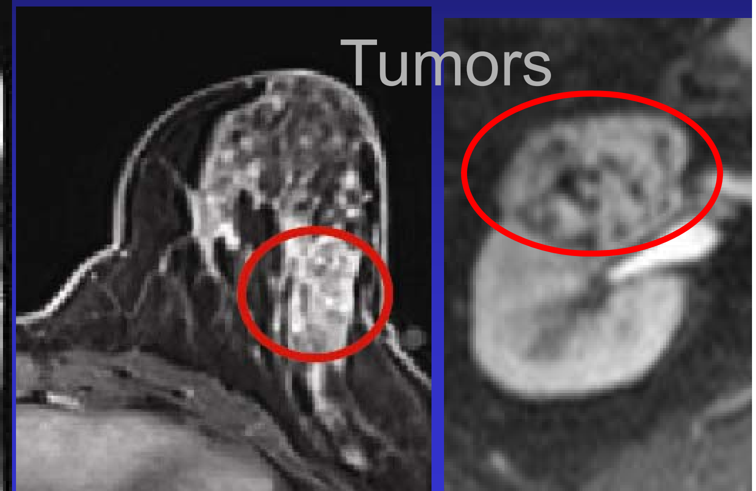
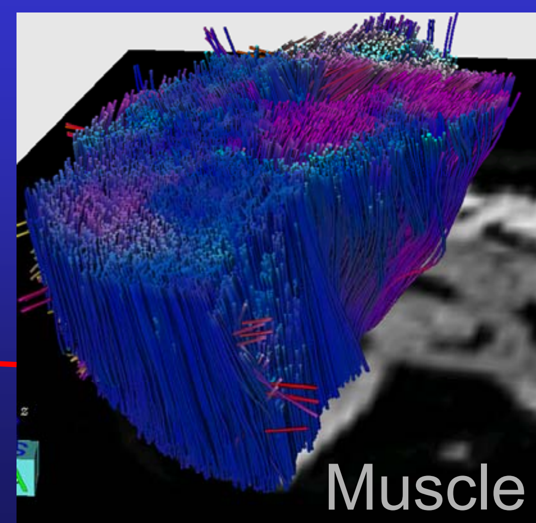
  - Common Artifacts
  - Signal-to-Noise Ratio (SNR)
- Diffusion MRI
- Functional MRI

DWI in the Human Body





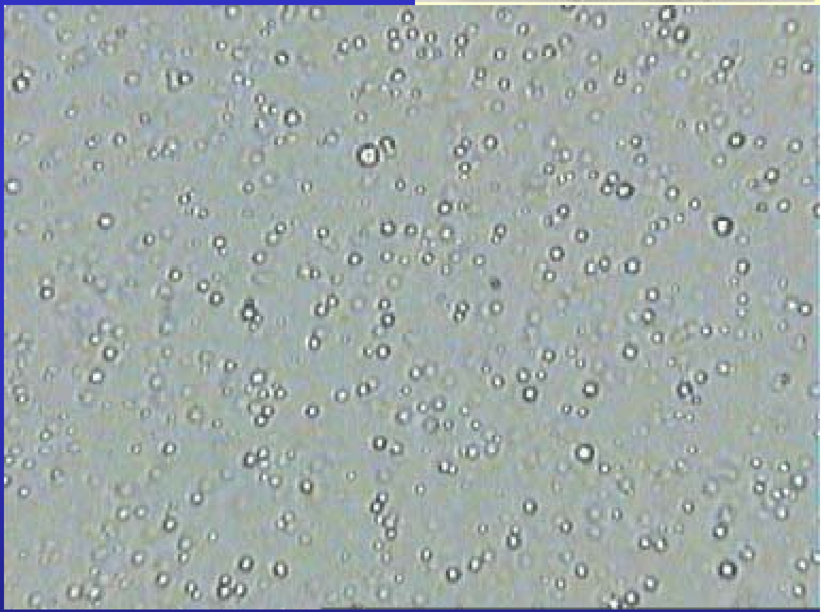


1. MR Diffusion Contrast

Brownian Motion

Robert Brown

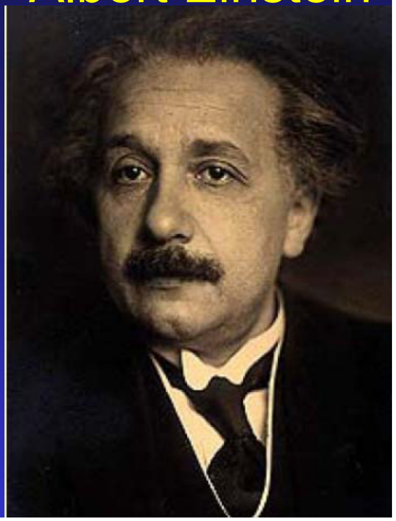
- First discovered 1828 (Brown)
- Random motion of molecules in liquid or gas
- Controlled by :
  - Molecular size / weight
  - Intermolecular forces (viscosity)
  - Temperature
  - Structure of confining medium



Fat globules in diluted milk

Diffusion Theory

Albert Einstein



- <u>Propagator</u>

P(x,t) = probability of displacement *x* in time *t*

- Free Diffusion equation

$$\frac{\partial P}{\partial t} = D\frac{\partial^2 P}{\partial x^2}$$

- Solution : Gaussian function

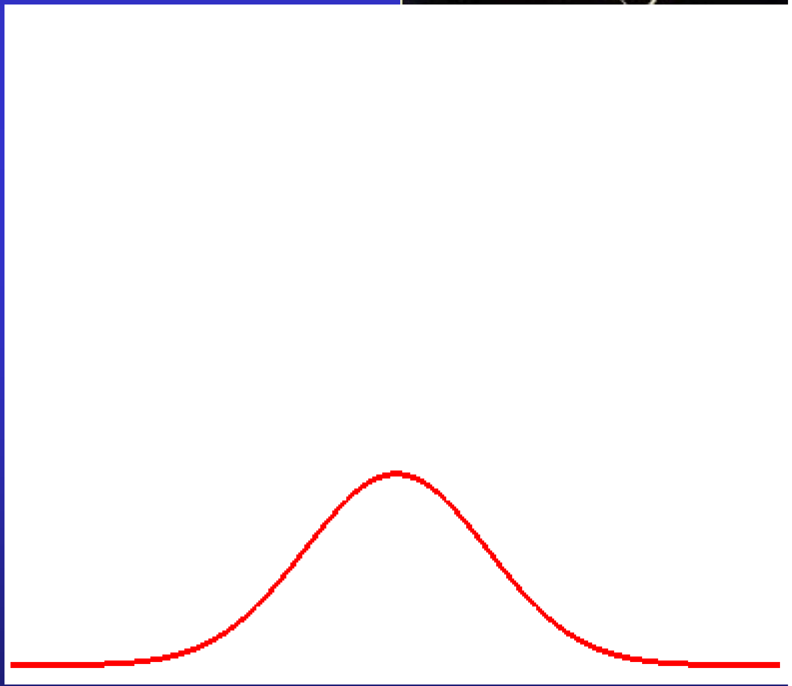
$$P(x,t) = \sqrt{\frac{1}{4\pi Dt}}\exp\left[-\frac{x^2}{4Dt}\right]$$

- Mean-squared displacement

$$\left\langle x^2 \right\rangle = \int x^2 P(x,t)\,dx = 2Dt$$

- Average diffusion length

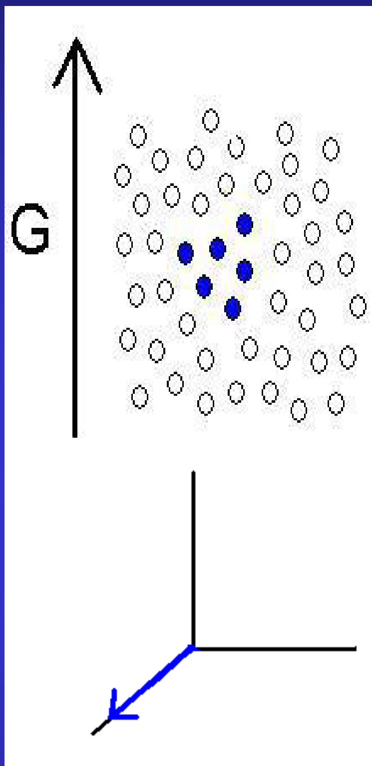
$$l_D \approx \sqrt{2Dt}$$

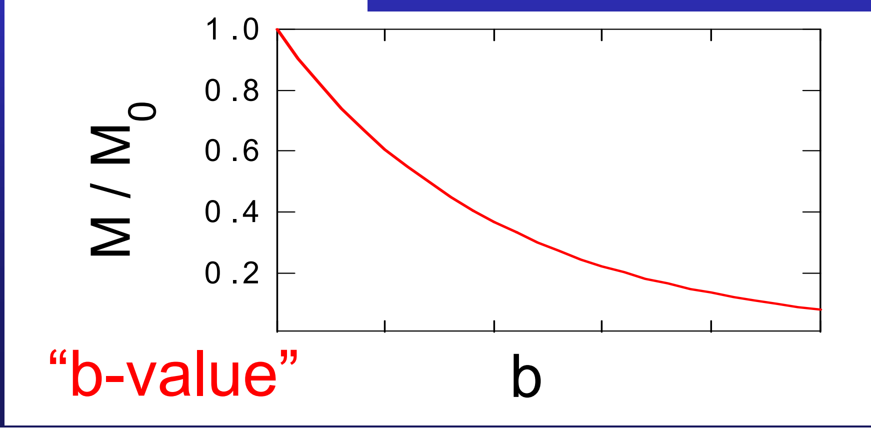


"Gaussian" Diffusion

MR Diffusion

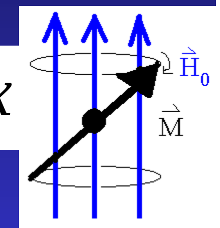
MR Diffusion Weighting





Precession in field gradient:

$$\omega_0 = \gamma H_0 = \gamma h_0 + \gamma G x$$



Spatial dispersion:

$$\langle \Delta x^2 \rangle = 2Dt$$

Phase dispersion:

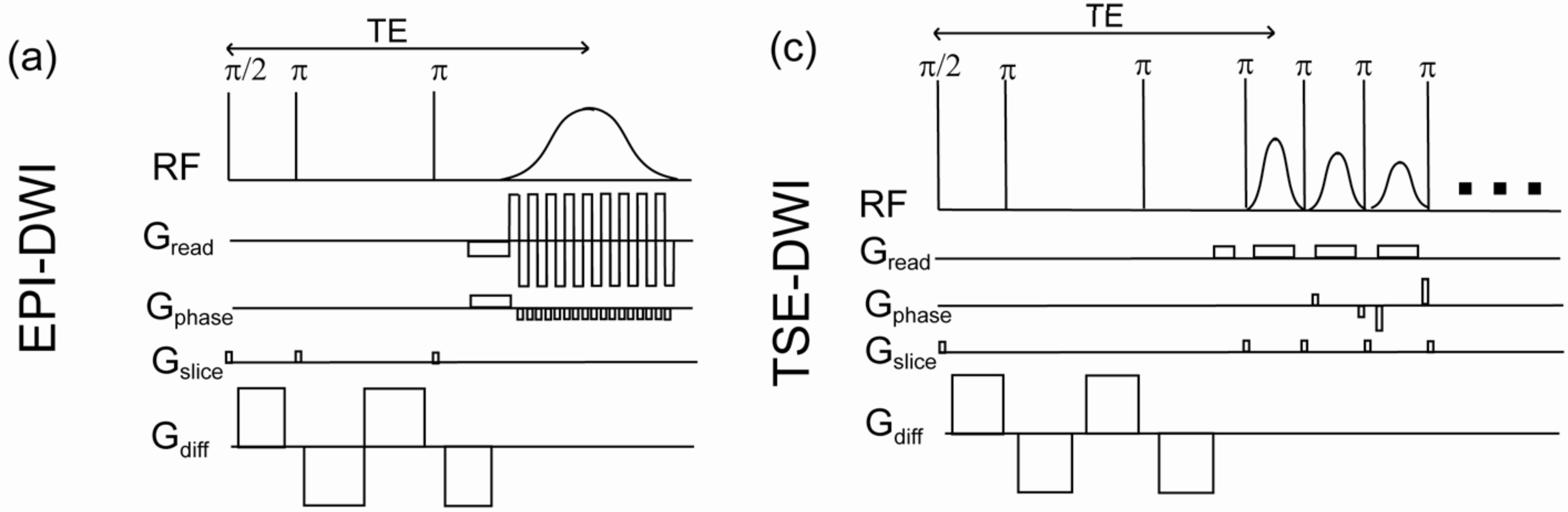
$$\langle \phi^2 \rangle \propto \gamma^2 G^2 D t^3$$

Magnetization:

$$\frac{M_t}{M_0} = \langle e^{i\phi} \rangle \approx e^{-\frac{\langle \phi^2 \rangle}{2}}$$

$$= \exp(-bD)$$

Clinical MRI pulse sequences



Twice-refocused spin echo, bipolar gradients

Echo-planar Imaging (EPI)

Turbo spin echo (TSE)

General b-value calculation

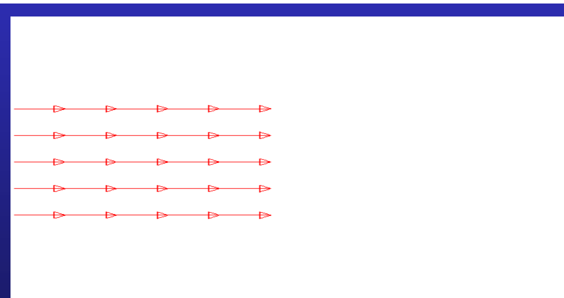
- For gaussian diffusion, can find closed form

$$b = \gamma^2 \int_0^{TE} \left[\int_0^t G_{eff}(t')\,dt' \right]^2 dt \equiv \int_0^{TE} k^2(t)\,dt$$

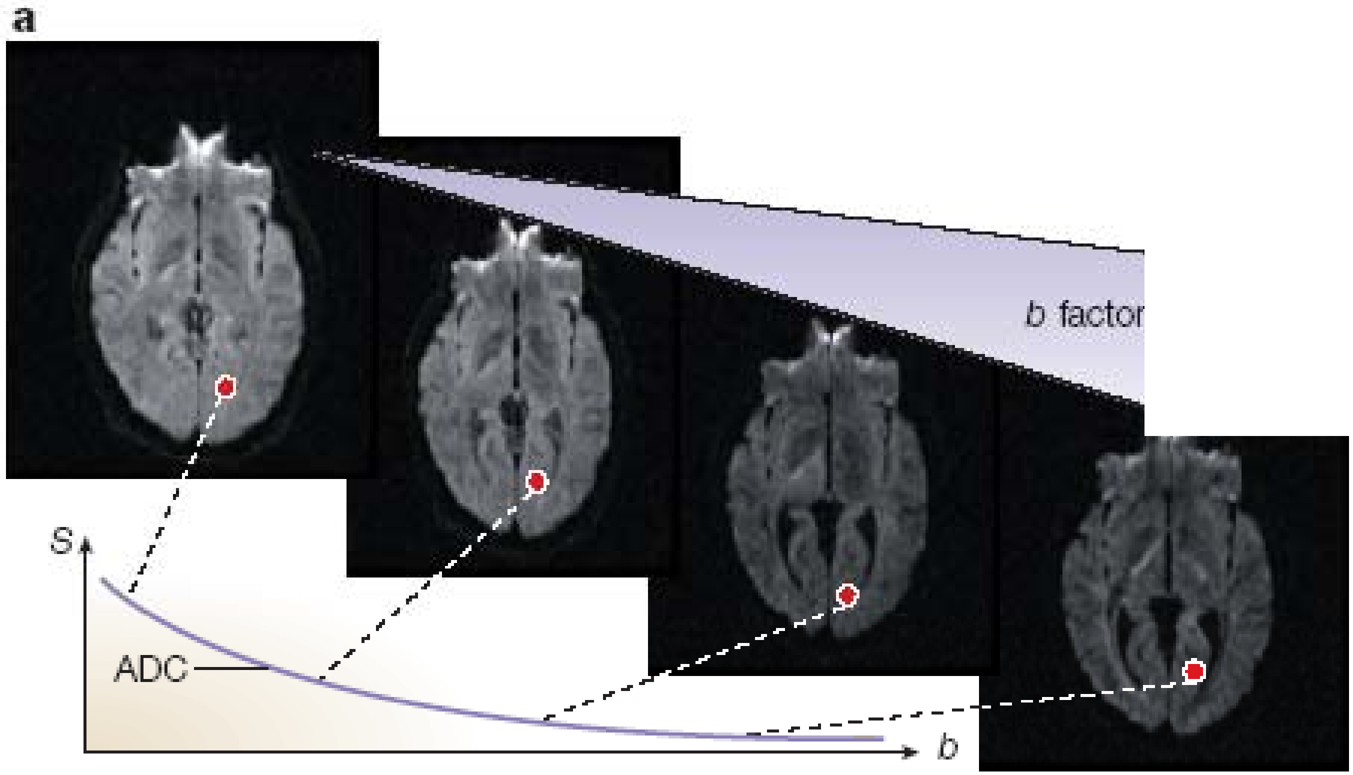
$G_{eff}(t')$: effective gradient (including RF inversions)

- Similarly, can calculate net phase shift due to constant velocity motion (e.g. flow) for any waveform

$$\phi = \vec{f} \cdot \vec{v} \qquad \vec{f} = \int_0^{TE} t' G_{eff}(t')\,dt' = -\int_0^{TE} k(t)\,dt$$



Diffusion-Weighted MRI (DWI)



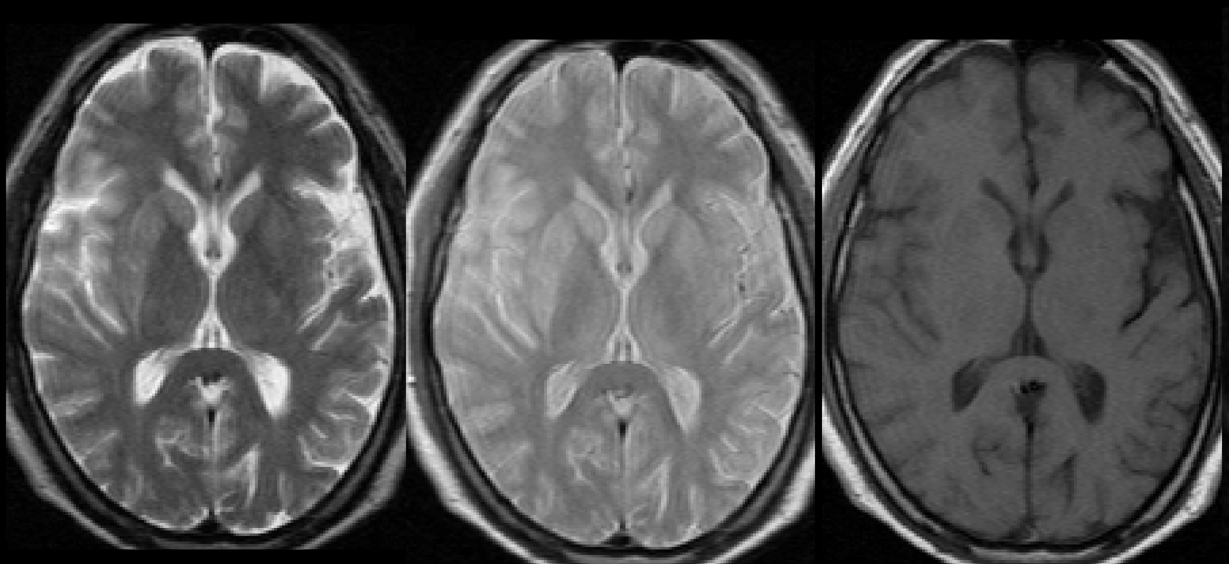
D. Le Bihan. Nature Reviews Neuroscience 2003;4:469-480

Diffusion-Weighted MRI (DWI)

- The degree to which the pulse sequence is sensitive to diffusion is expressed through the “b-value” or “b-factor”

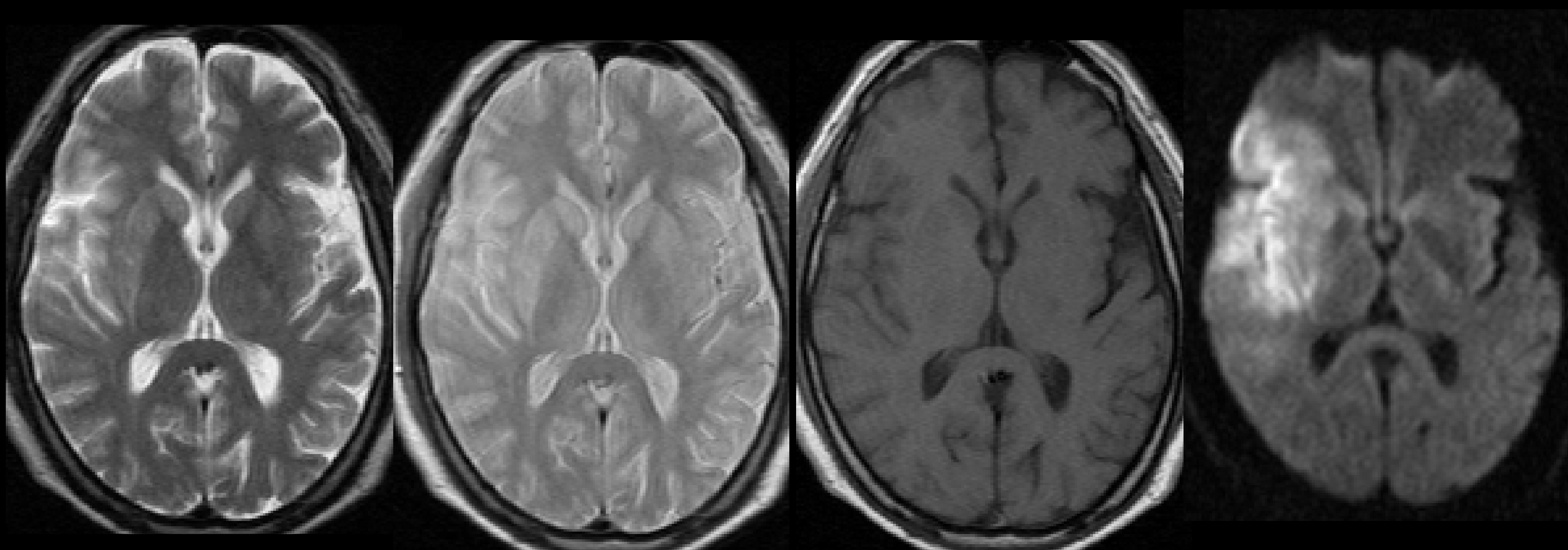
- The signal intensity of diffusion weighted images are
  – inversely related to the b-value
  – directly related to the degree of diffusion restriction (at a particular b-value)
  – directly related to T2

MRI of Acute Stroke



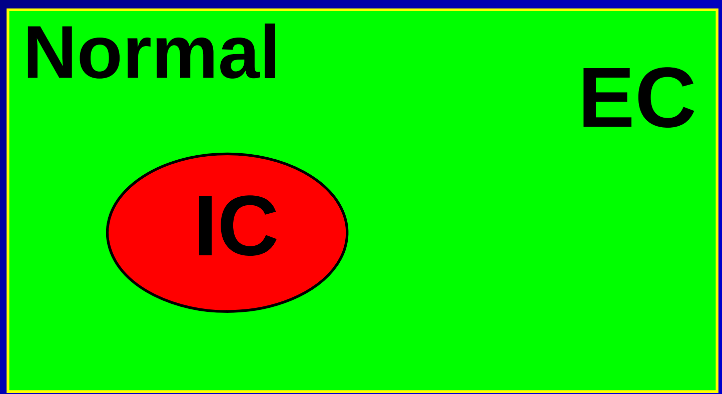
T2w PDw T1w

MRI of Acute Stroke

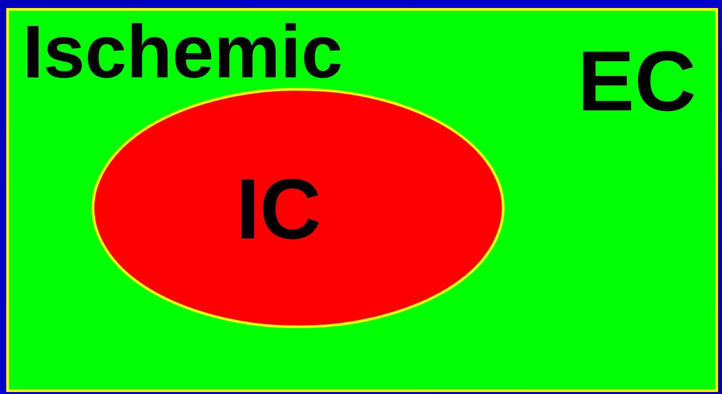


T2w PDw T1w DWI (b = 1000)

Cell Swelling Hypothesis



Intracellular diffusion is slower than extracellular diffusion



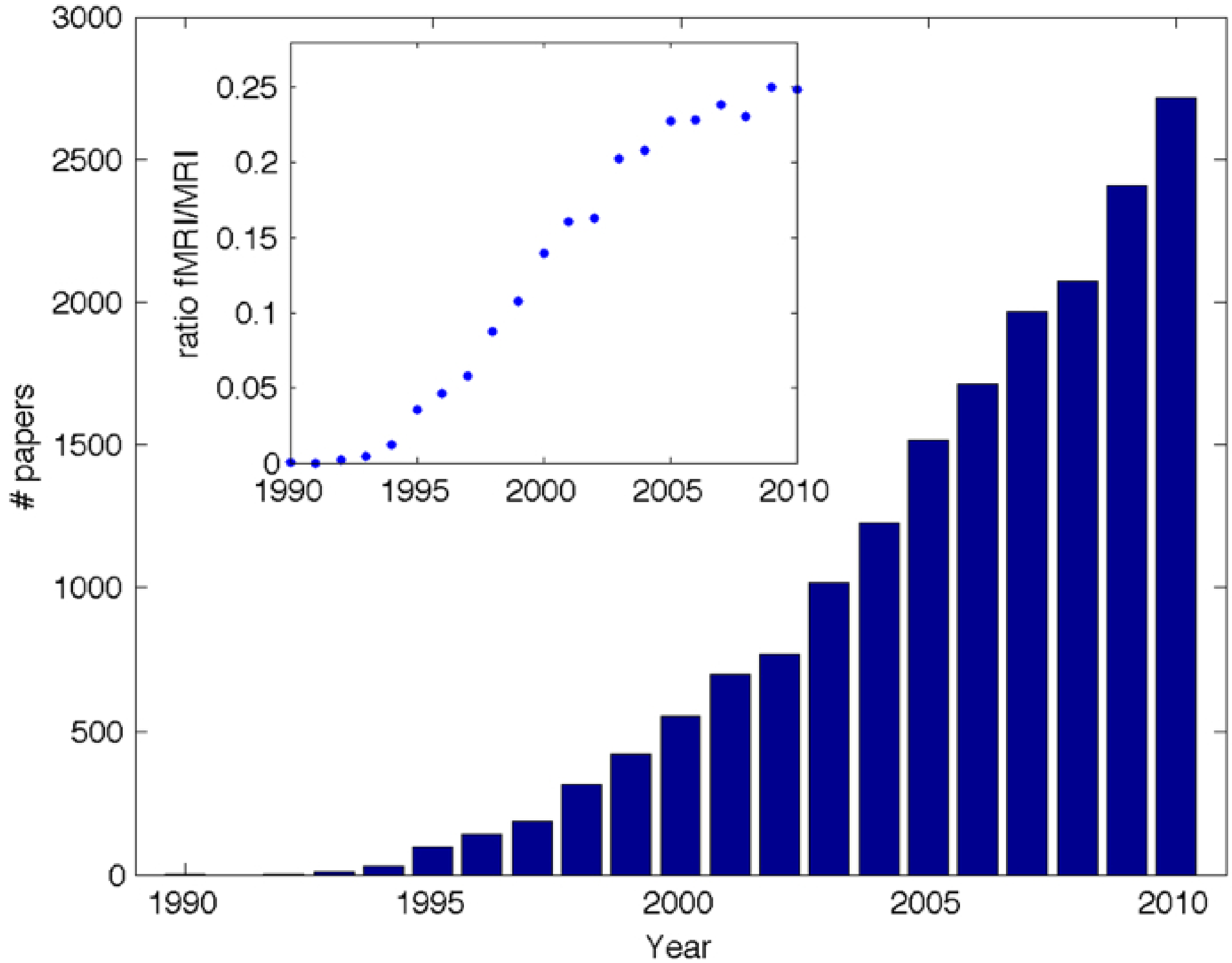
Swollen intracellular volume
Increased volume average DWI

Outline

- Review of Image Reconstruction
- Image Quality
  - Image Resolution
  - Common Artifacts
  - Signal-to-Noise Ratio (SNR)
- Diffusion MRI
- Functional MRI

Importance of fMRI

Number of publications per year in PubMed with “**fMRI**” or “**functional MRI**” in the title or abstract



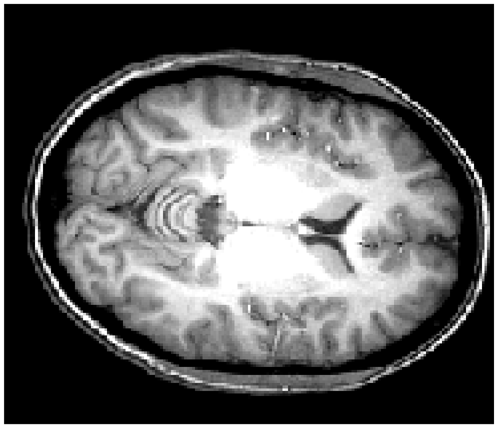
Overview

- Used to determine which area of brain is responsible for a given cognitive task
- The oxygenation status of the blood stream changes after the onset of a neuron activity
- fMRI: Taking a time series of MRI images of the brain after the subject is given a cognitive task
  - At each voxel, plot the MRI signal in time
  - Neuron activity at a voxel usually leads to a special MRI signal pattern
  - Detect such patterns
  - Voxels with detected patterns are considered to be responsible for the cognitive task

MRI vs. fMRI

MRI

high resolution
(1 mm)

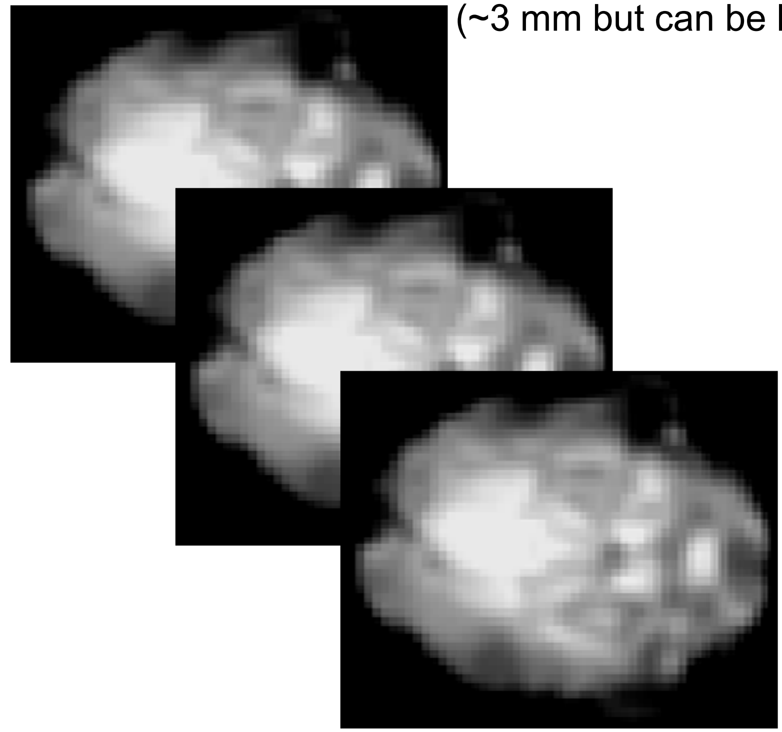


one image

http://www.fmri4newbies.com/

fMRI

low resolution
(~3 mm but can be better)

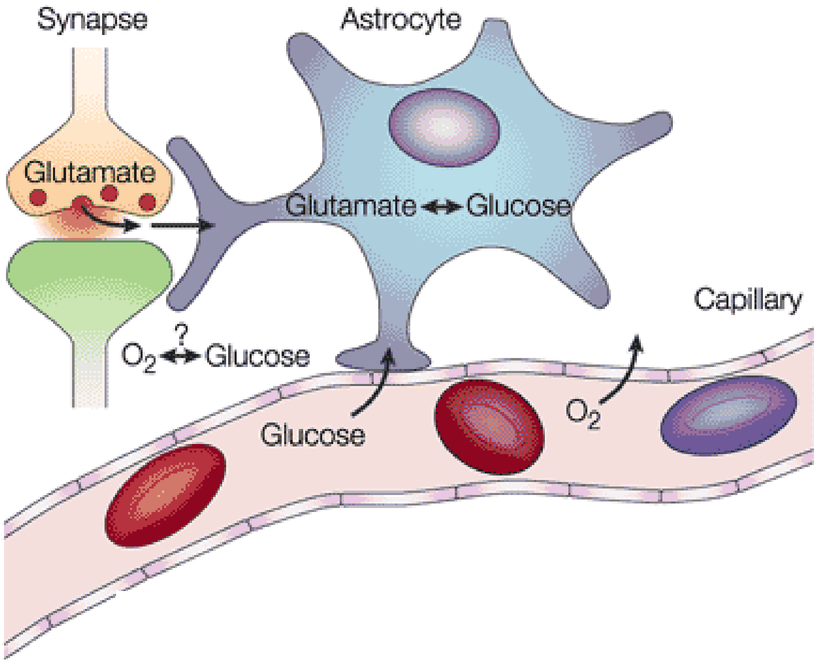


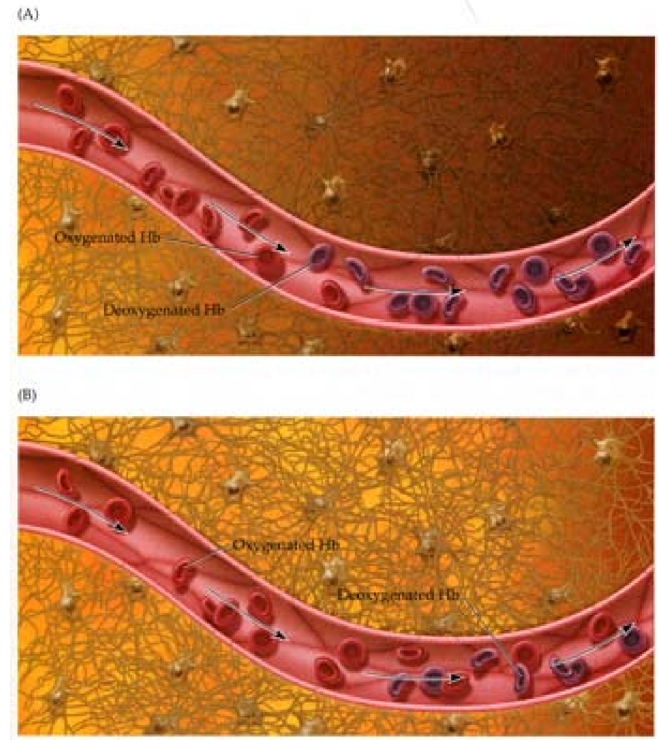
…

many images
(e.g., every 2 sec for 5 mins)

↑ neural activity ➔ ↑ blood oxygen ➔ ↑ fMRI signal

A basic description of BOLD fMRI physics





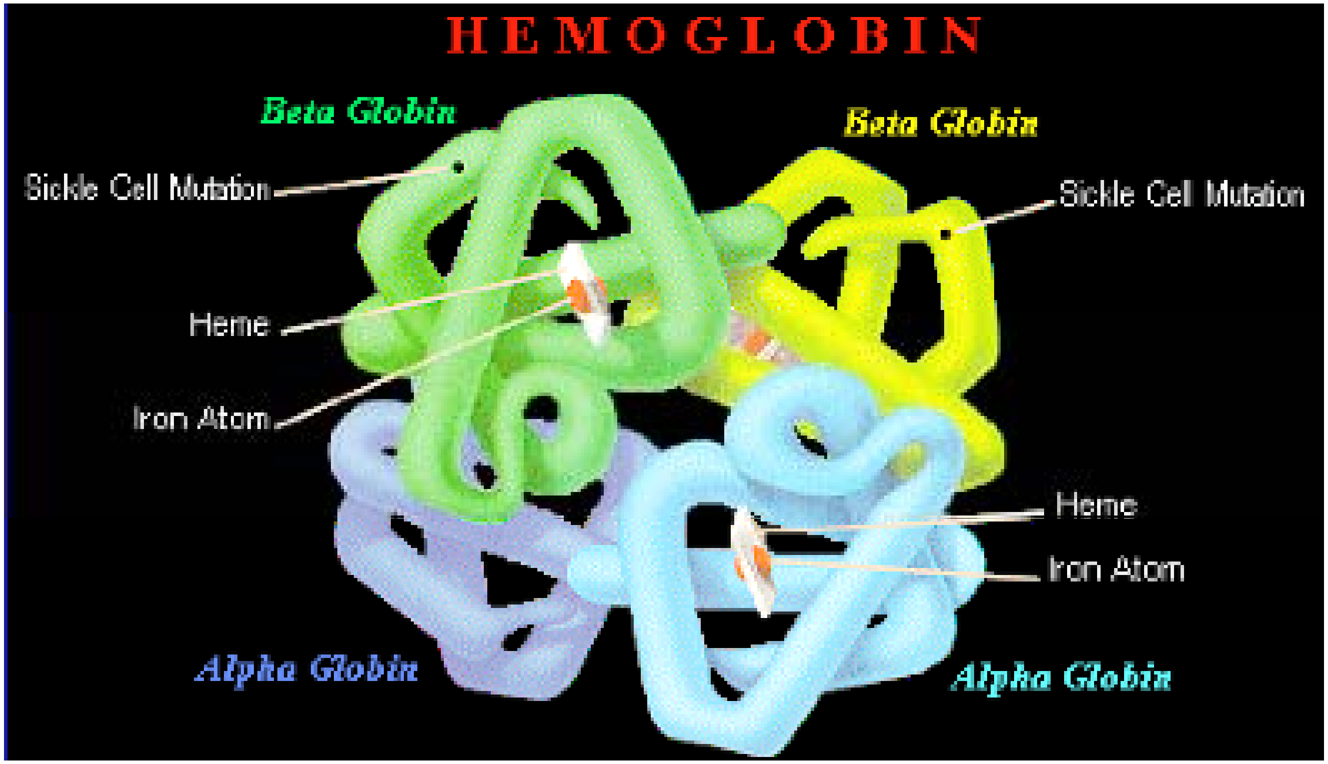
Baseline

Task

Increased neural activity leads to increased blood flow, blood volume, and oxygen consumption

Roy and Sherrington (1890)
without the pretty graphics

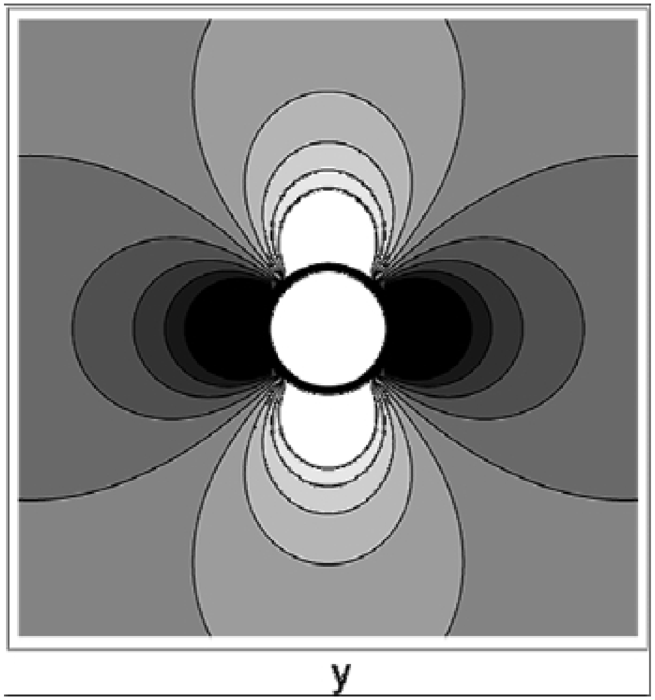
Contrast agents?

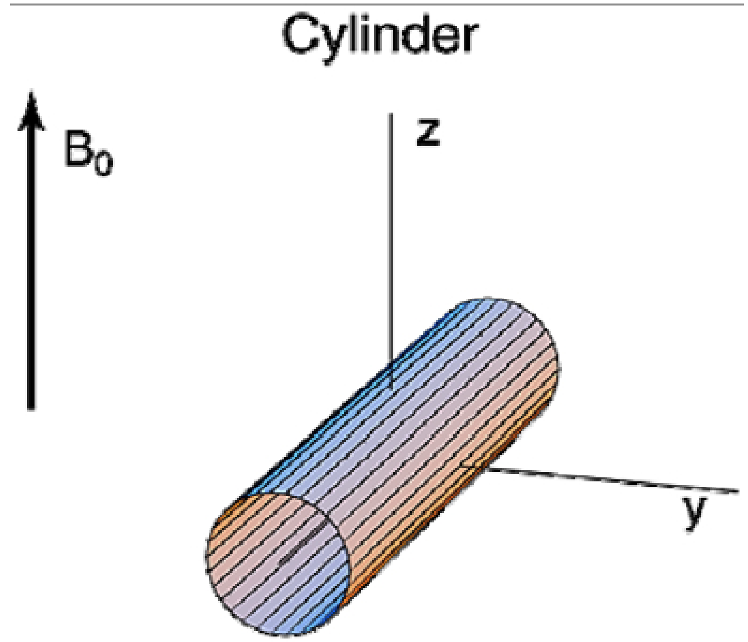


4 Iron atoms Bind O_2

Oxy-hemoglobin: diamagnetic
Deoxyhemoglobin: paramagnetic
Changes local magnetic field

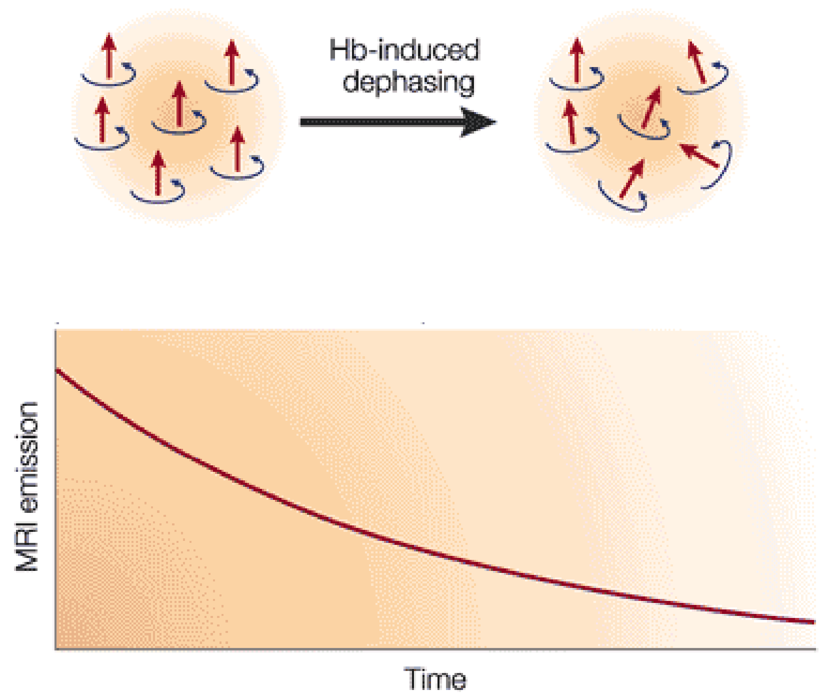
Magnetic Field Near a Vessel

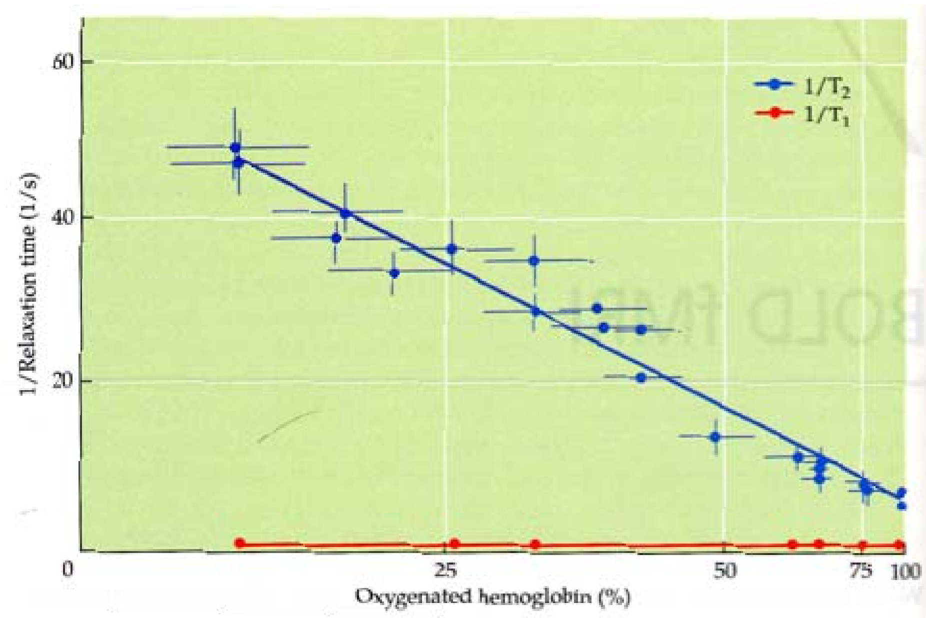




Field depends on several things:

1) Location
2) Vessel orientation relative to B_0
3) Deoxyhemoglobin content





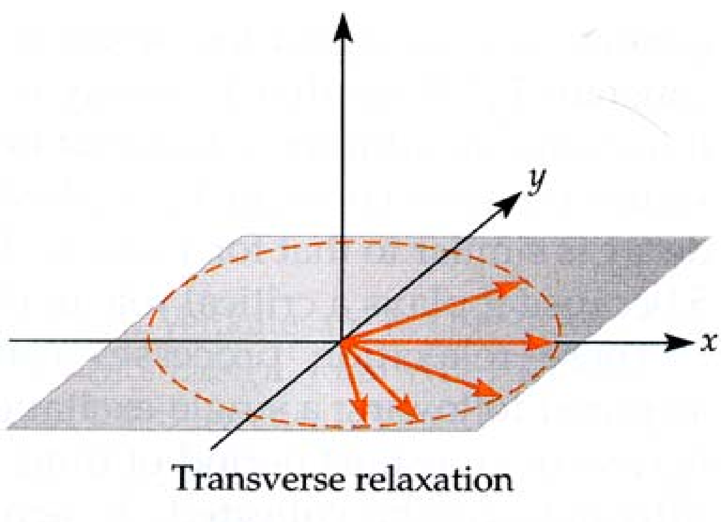
Thulborn 1982

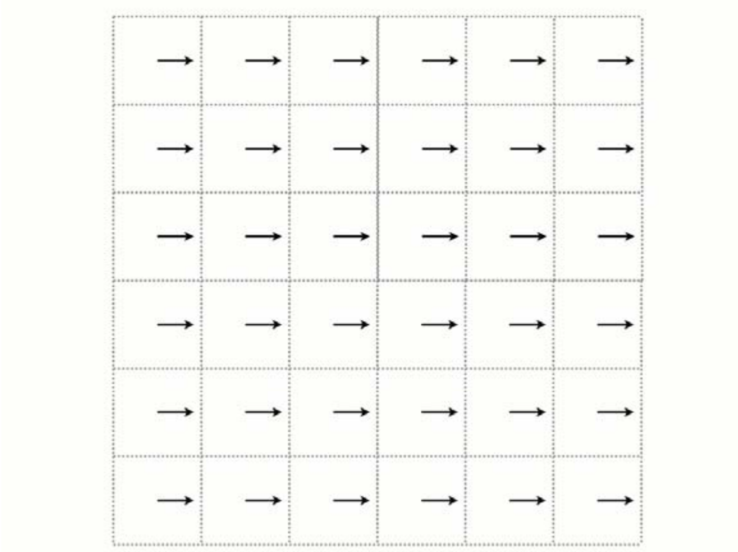
Oxygenation of hemoglobin changes local magnetic field and T_2 of blood

Inhomogeneous field $\Rightarrow$ different phases $\Rightarrow$ de-phasing

T_2^* Decay

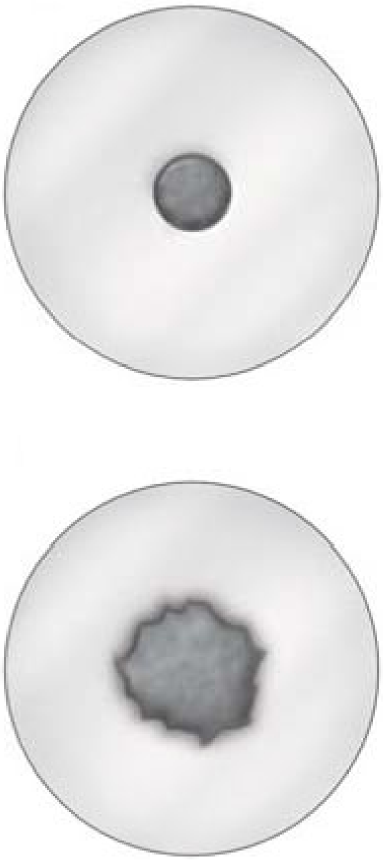
Due to variation of magnetic field INSIDE in a voxel





Deoxyhemoglobin in veins changes T_2^*

Deoxygenated Blood → Signal Loss

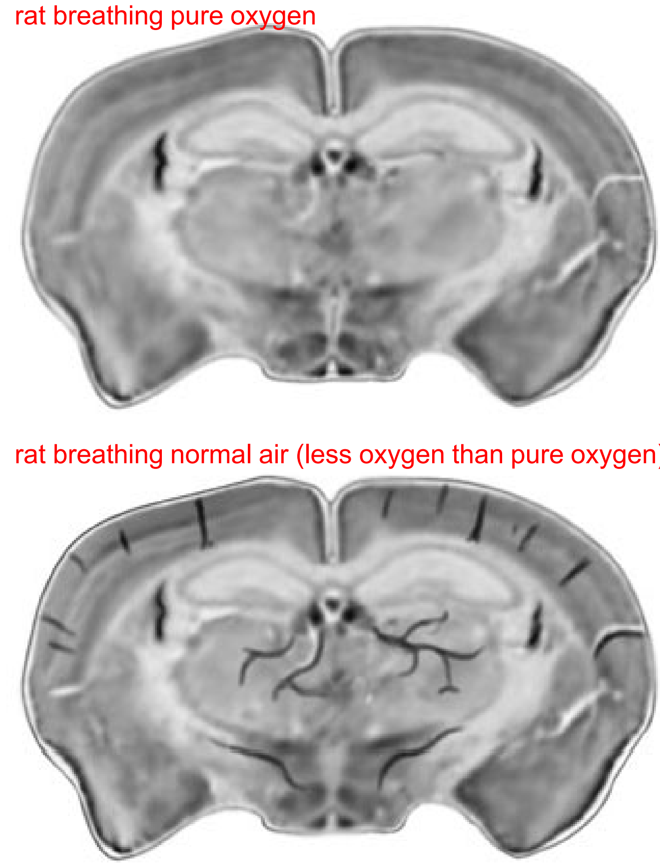


Oxygenated blood?

- Diamagnetic
- Doesn't distort surrounding magnetic field
- No signal loss…

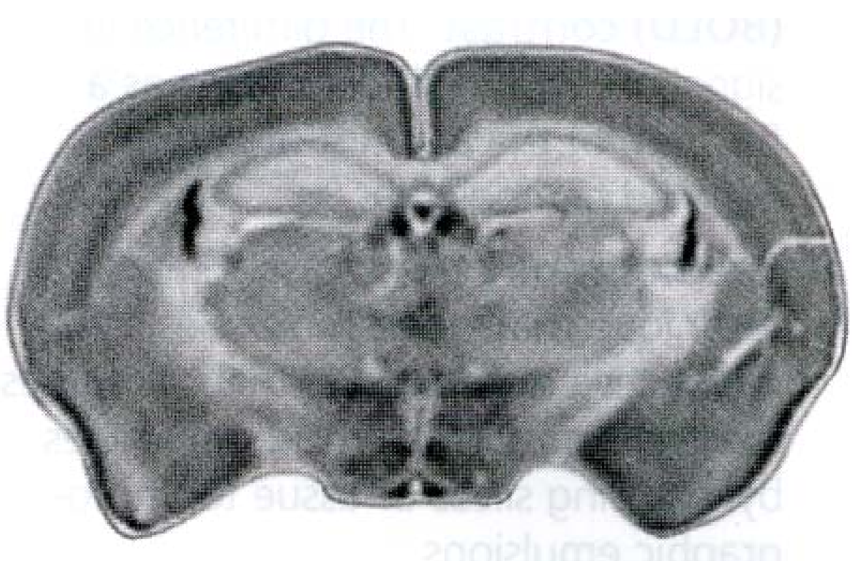
<u>De</u>oxygenated blood?

- Paramagnetic
- Distorts surrounding magnetic field
- Signal loss !!!

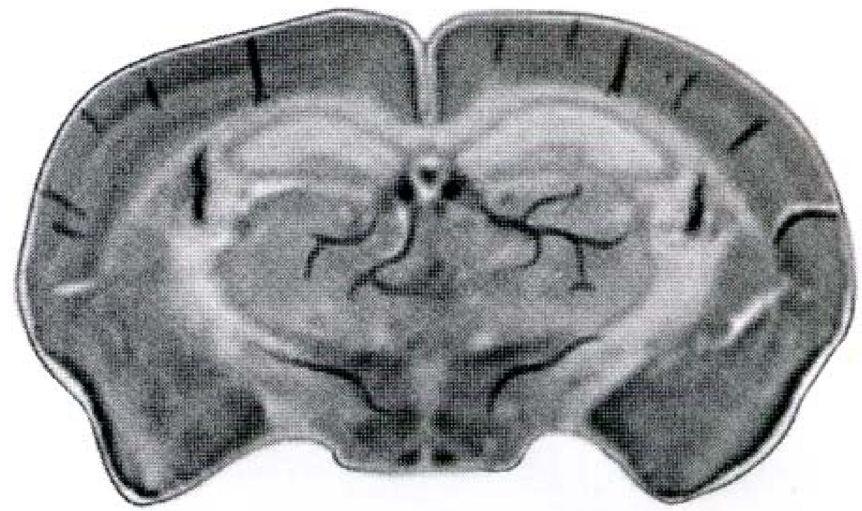


Images from Huettel, Song & McCarthy, 2004, Functional Magnetic Resonance Imaging based on two papers from Ogawa et al., 1990, both in Magnetic Resonance in Medicine

The BOLD Effect



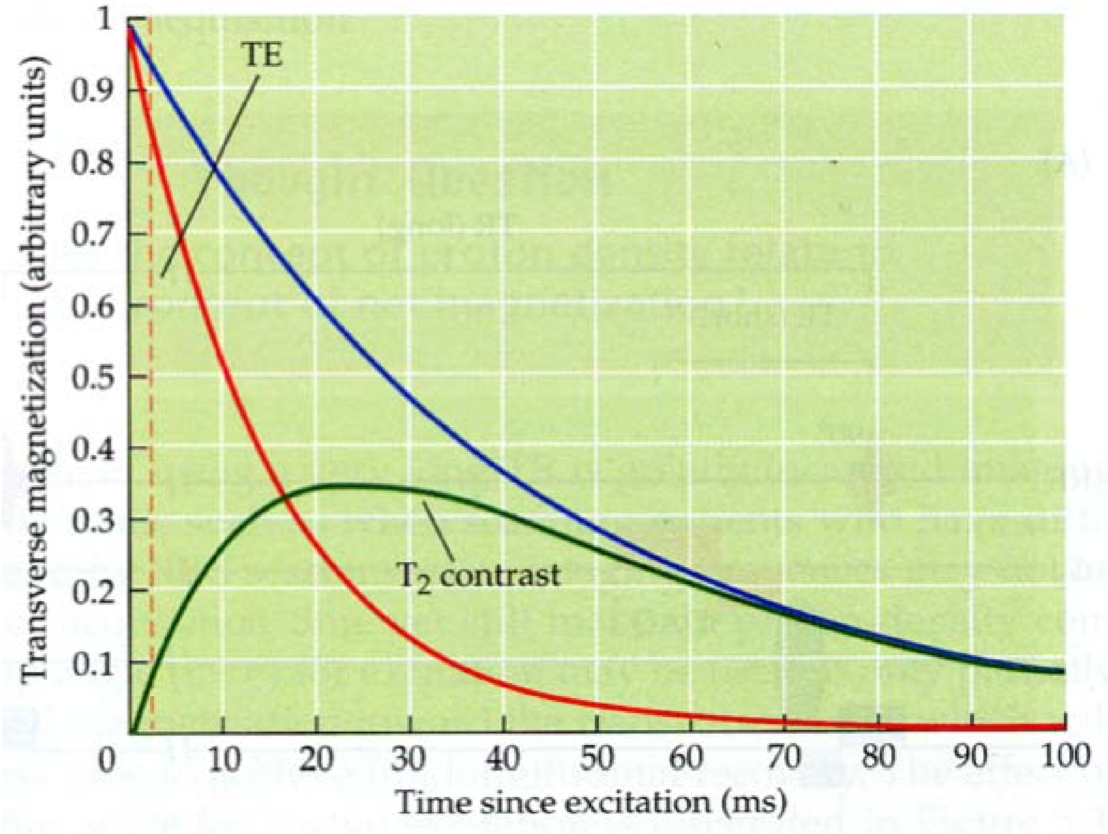
Pure O_2

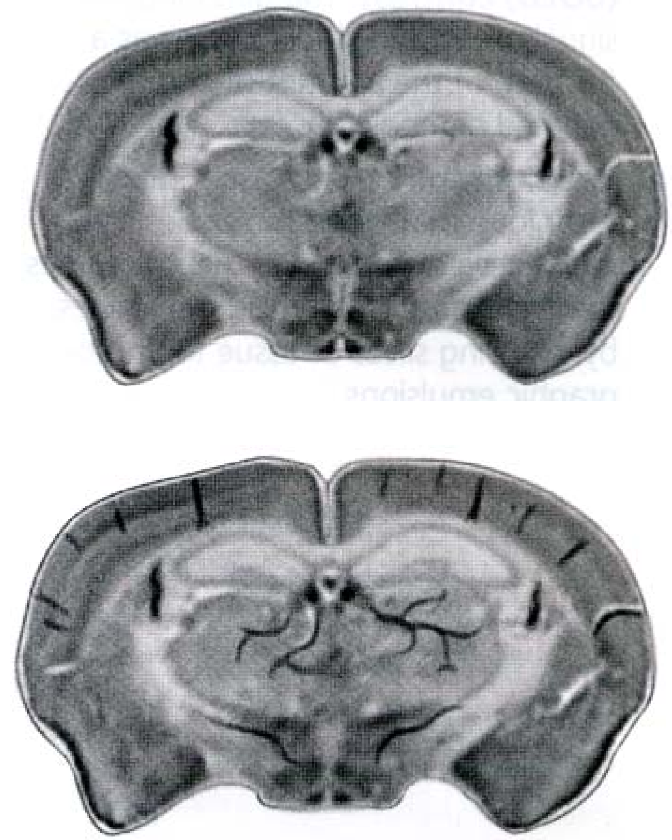


Normal Air
(21% O_2)

Oxygenation of blood can be imaged!
Ogawa 1990

T_2^* Image Contrast





Pick TE to maximize T_2^* sensitivity

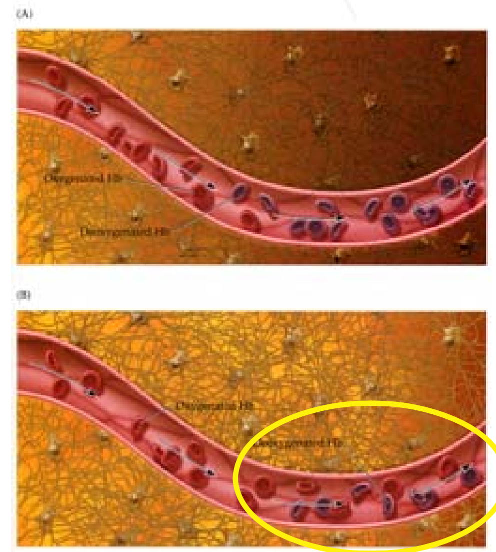
Neuronal activation

↓

Local hemodynamic changes

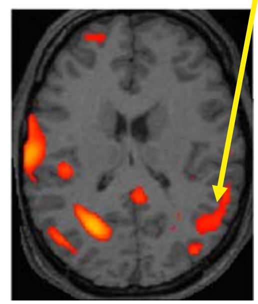
- ↑ Blood flow
- ↑ Blood volume
- ↑ oxygen consumption

↓



Baseline

Task



Kwong ‘92
Ogawa ‘92

Decrease in venous deoxyHb concentration → Local increase in MR signal

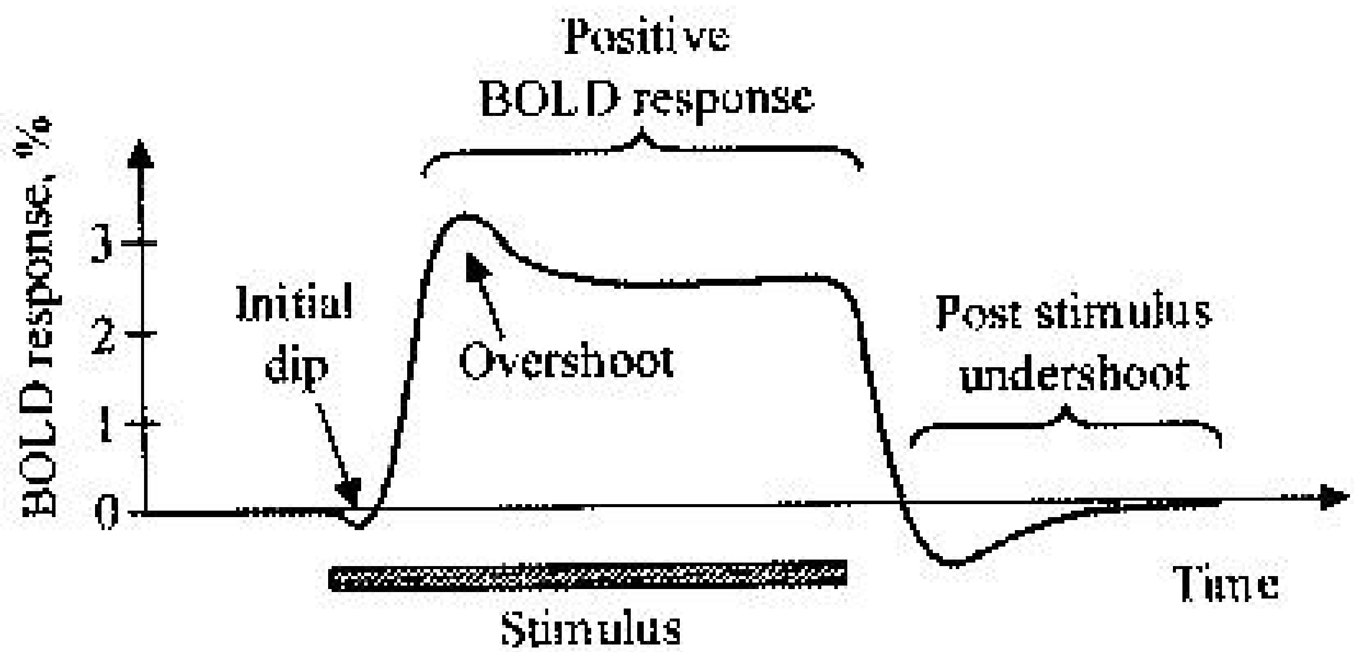
Changes of the BOLD fMRI Signal

- The function of the BOLD fMRI signal against time in response to a temporary increase in neuronal activity is known as the hemodynamic response function (HRF)
- After the onset of a neuron activity, the active neurons use oxygen thereby increasing the relative level of deoxyhaemoglobin in the blood, which leads to the decrease of the BOLD fMRI signal initially.
- Following this initial increase in deoxyhaemoglobin, there is a massive oversupply of oxygen-rich blood (reaching maximum at ~6s), leading to a large decrease in deoxyhaemoglobin, and hence increase in the BOLD fMRI signal.
- Finally, the level of deoxyhaemoglobin slowly returns to normal and the BOLD fMRI signal decays until it has reached its original baseline level (~24s).

Based on: http://www.sph.sc.edu/comd/rorden/fmri_guide/fmri_guide.pdf

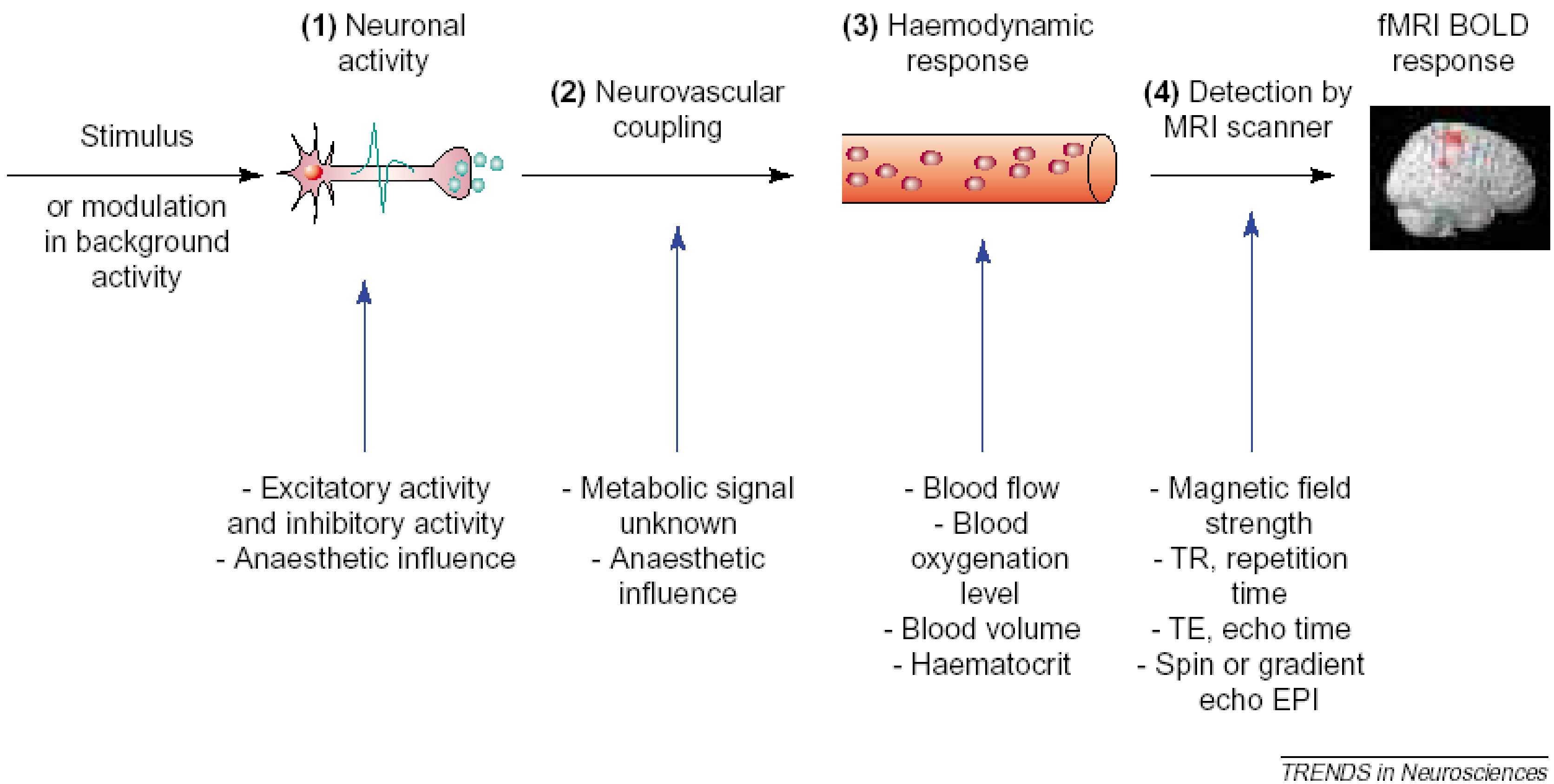
Hemodynamic Response Function (HEF)

- HEF is also known as BOLD Time course



From: http://www.fmri4newbies.com/

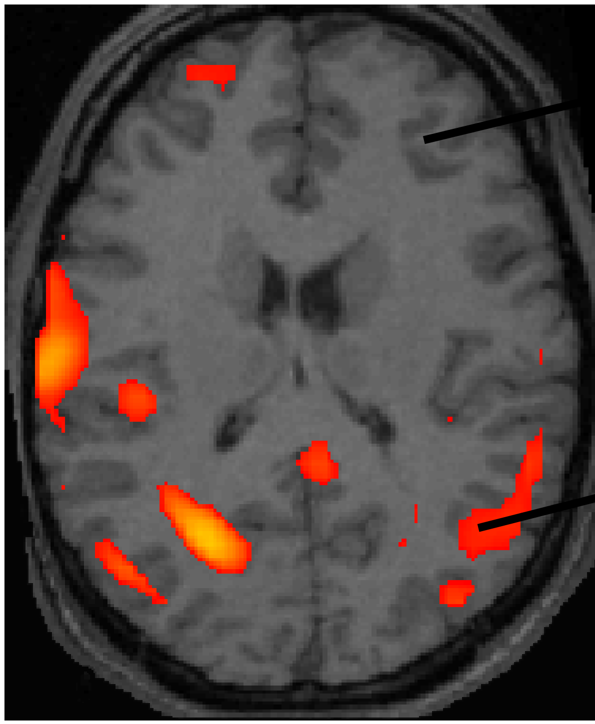
Stimulus to BOLD



Source: Arthurs & Boniface, 2002, *Trends in Neurosciences*

From: http://www.fmri4newbies.com/

Measure Cognitive Function

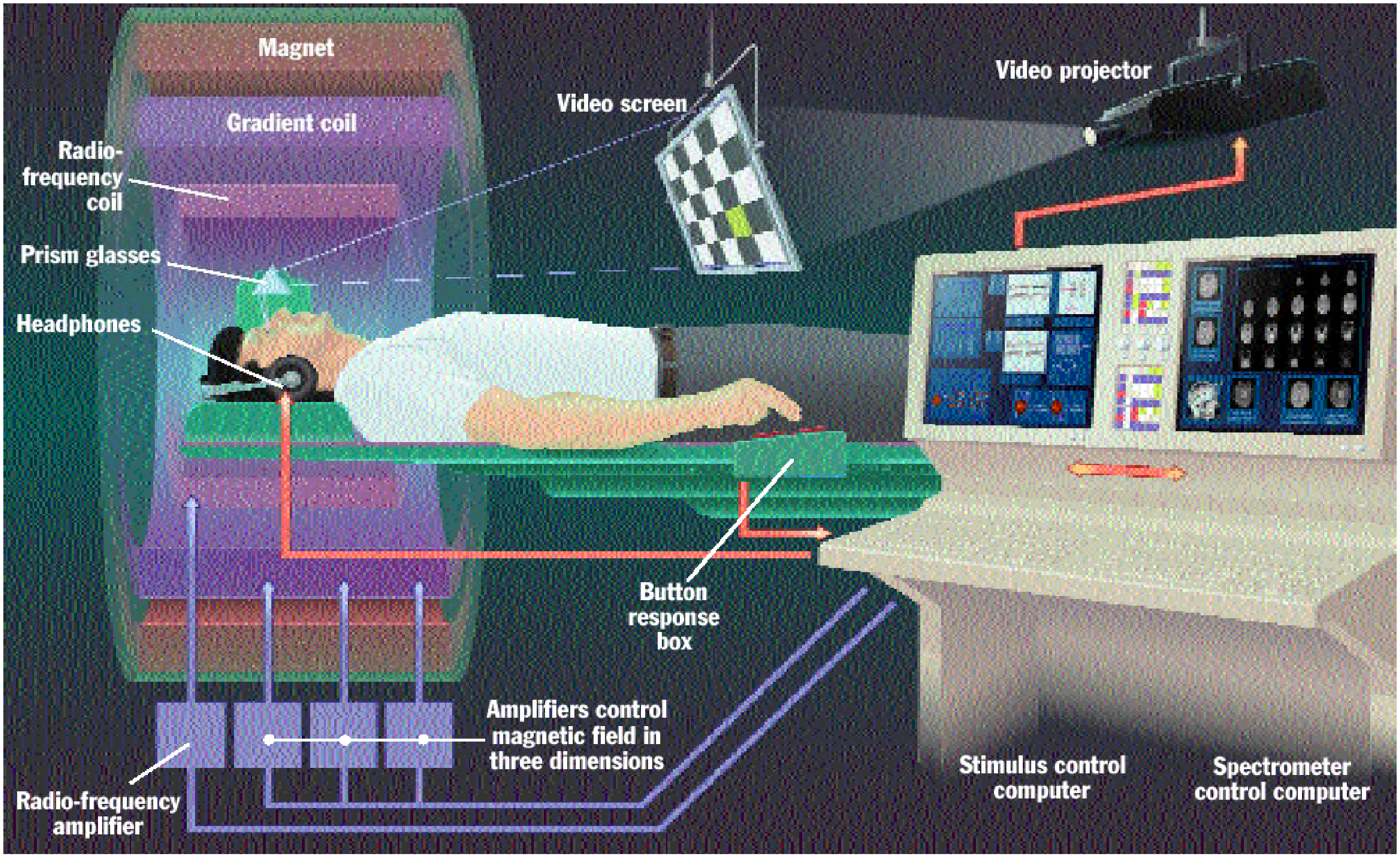


Anatomy image (T_1)

Statistical image overlay: color ~ P value

BOLD FMRI at 1.5T

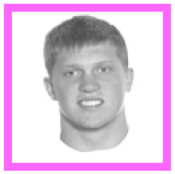
fMRI Set up



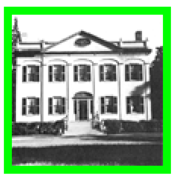
From: http://www.fmri4newbies.com/

Category-Specific Visual Areas

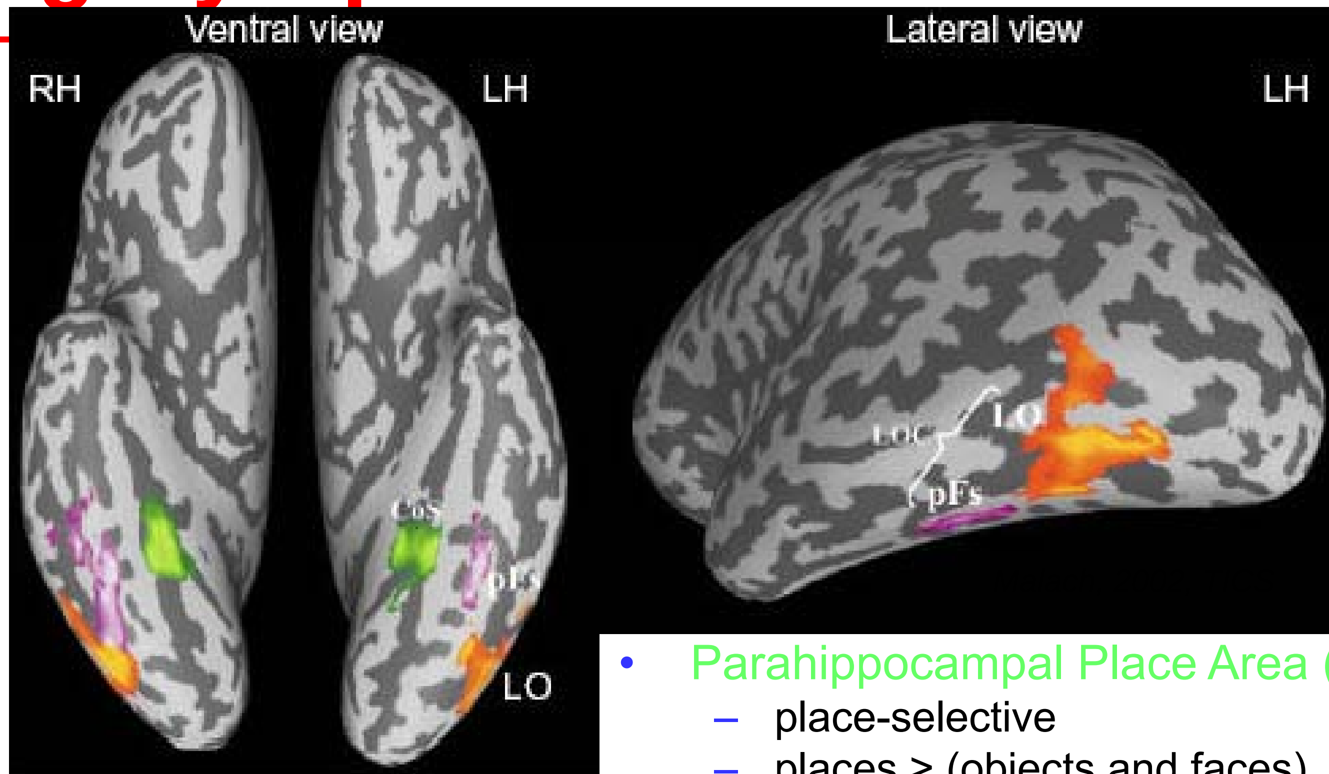
objects



faces



places



- Parahippocampal Place Area (PPA)
  - place-selective
  - places > (objects and faces)
  - places > scrambled images

- Lateral Occipital (LO)
  - object-selective
  - objects > (faces & scenes)
  - objects > scrambled images

- Fusiform Face Area (FFA) or pFs
  - face-selective
  - faces > (objects & scenes)
  - faces > scrambled images
  - ~ posterior fusiform sulcus (pFs)

http://www.fmri4newbies.com/

fMRI Experiment Stages: Prep

1) Prepare subject

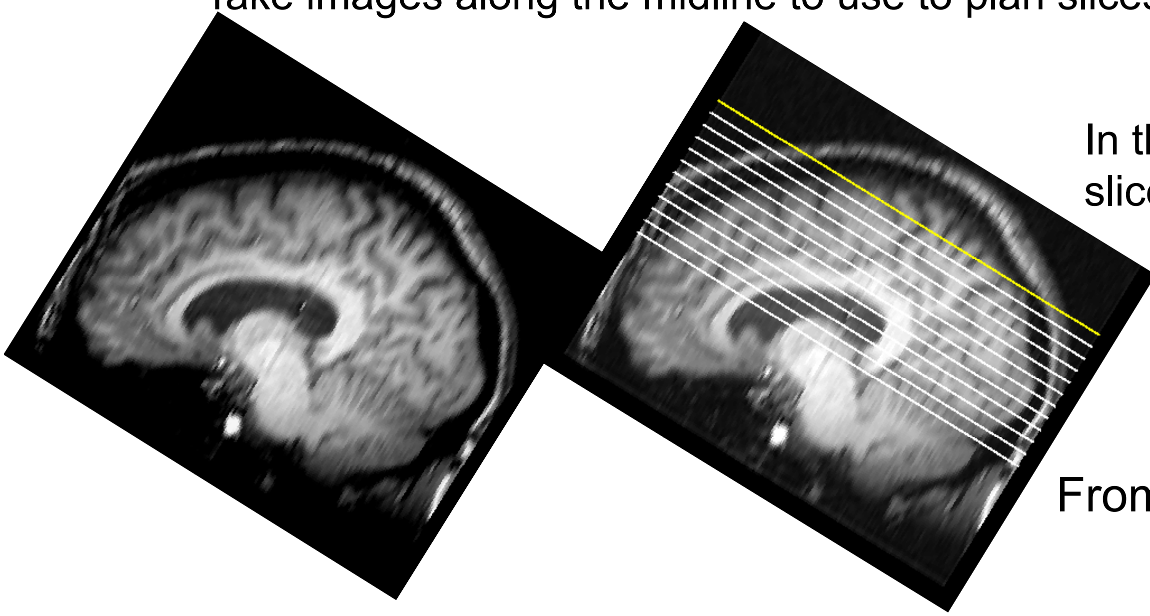
- Consent form
- Safety screening
- Instructions and practice trials if appropriate

2) Shimming

- putting body in magnetic field makes it non-uniform
- adjust 3 orthogonal weak magnets to make magnetic field as homogenous as possible

3) Sagittals

Take images along the midline to use to plan slices



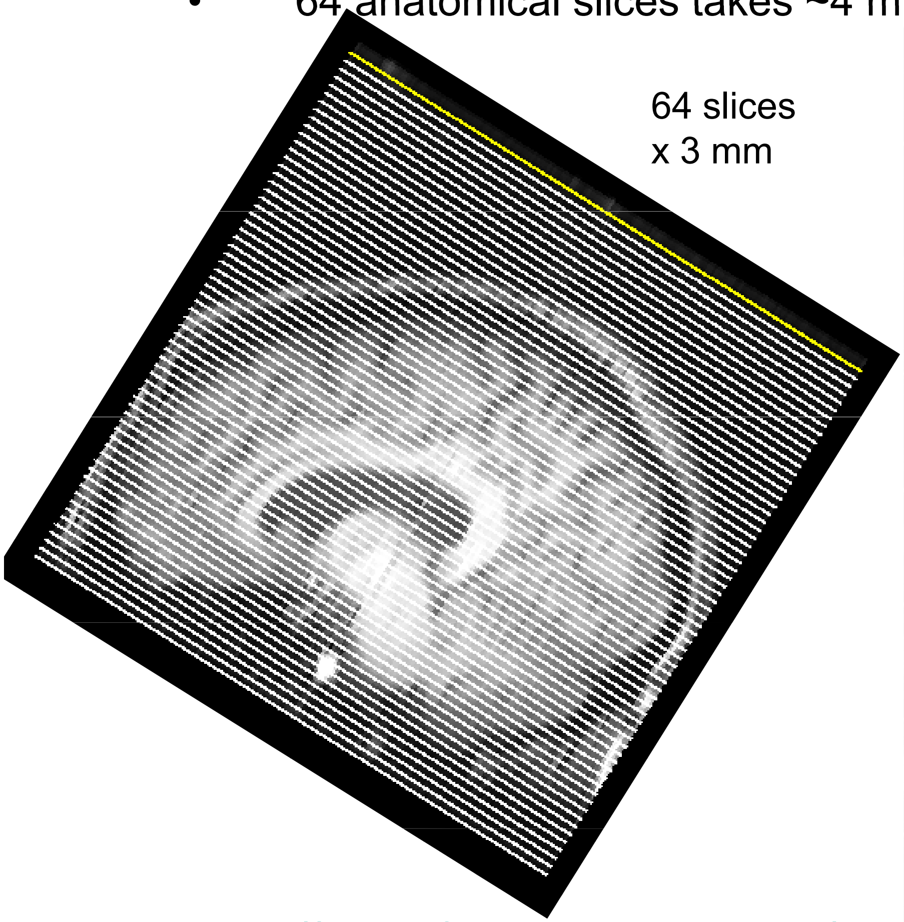
In this example, these are the *functional* slices we want: 12 slices x 6 mm

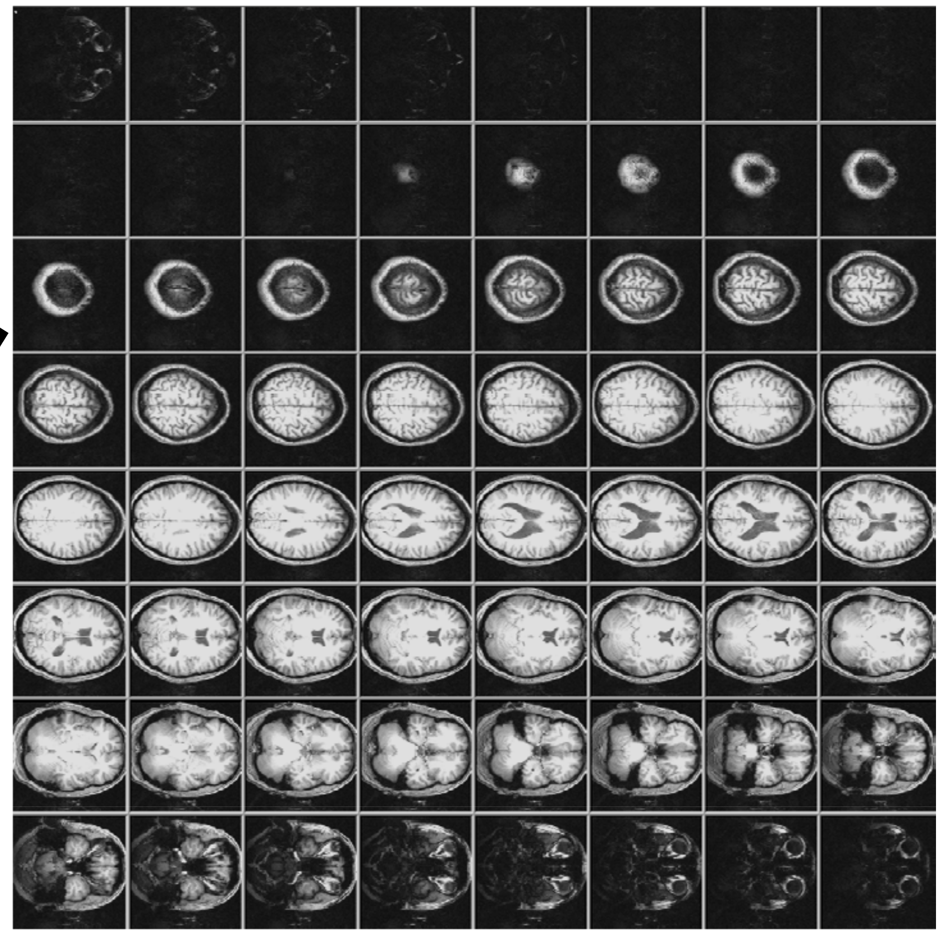
From: http://www.fmri4newbies.com/

fMRI Experiment Stages: Anatomicals

4) Take anatomical (T1) images

- high-resolution images (e.g., 0.75 x 0.75 x 3.0 mm)
- 3D data: 3 spatial dimensions, sampled at one point in time
- 64 anatomical slices takes ~4 minutes



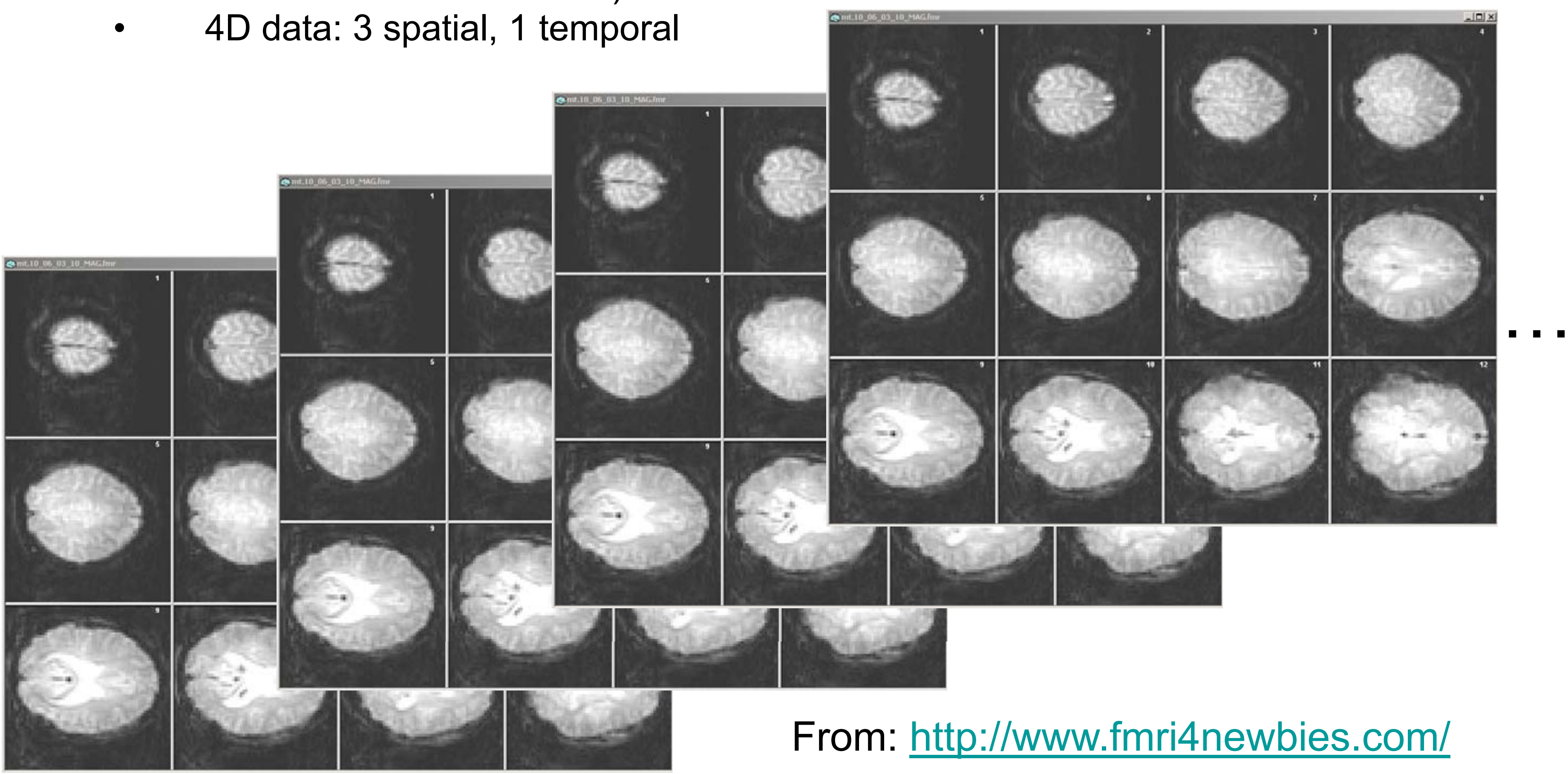


From: http://www.fmri4newbies.com/

fMRI Experiment Stages: Functionals

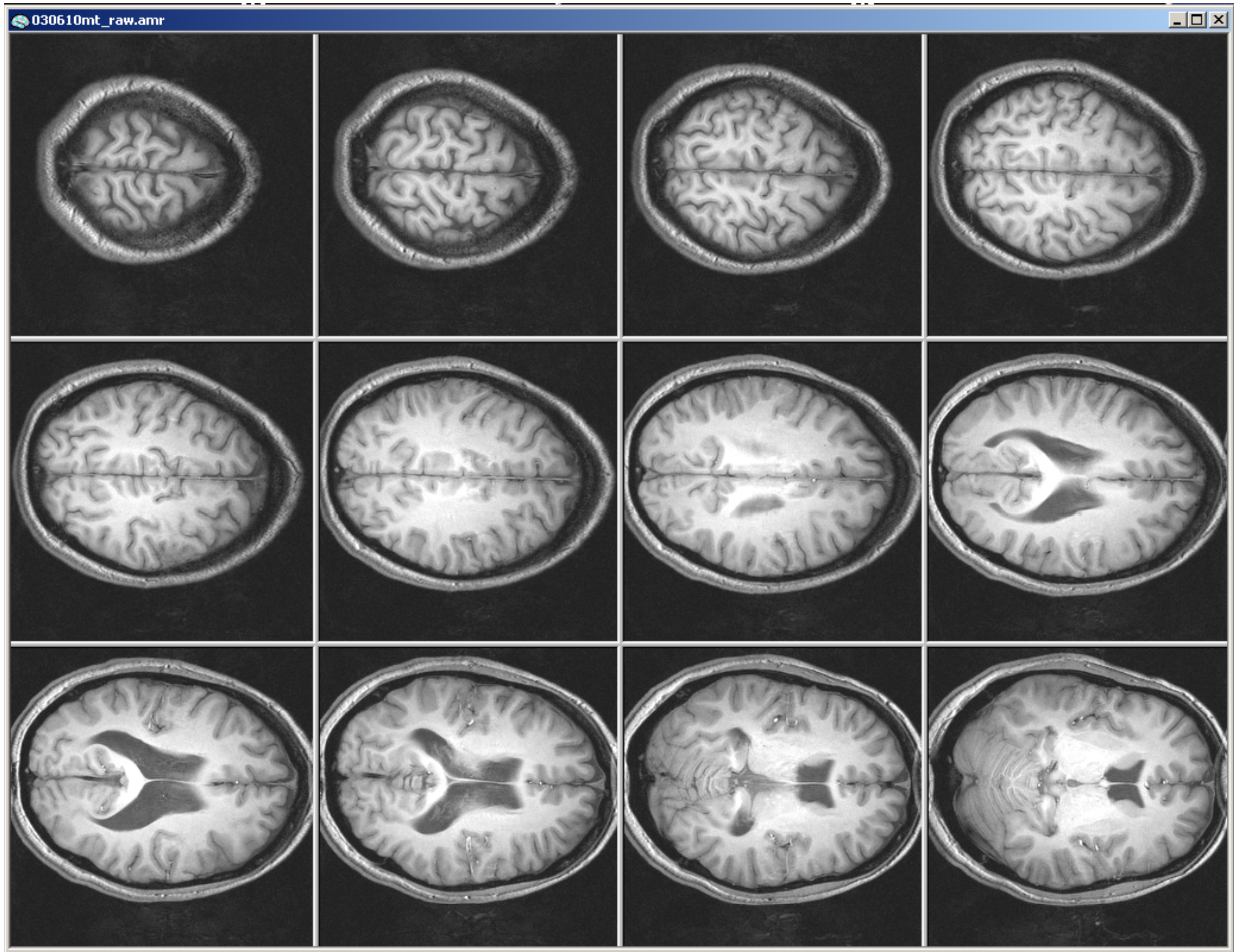
5) Take functional (T2*) images

- images are indirectly related to neural activity
- usually low resolution images (3 x 3 x 6 mm)
- all slices at one time = a volume (sometimes also called an image)
- sample many volumes (time points) (e.g., 1 volume every 2 seconds for 136 volumes = 272 sec = 4:32)
- 4D data: 3 spatial, 1 temporal



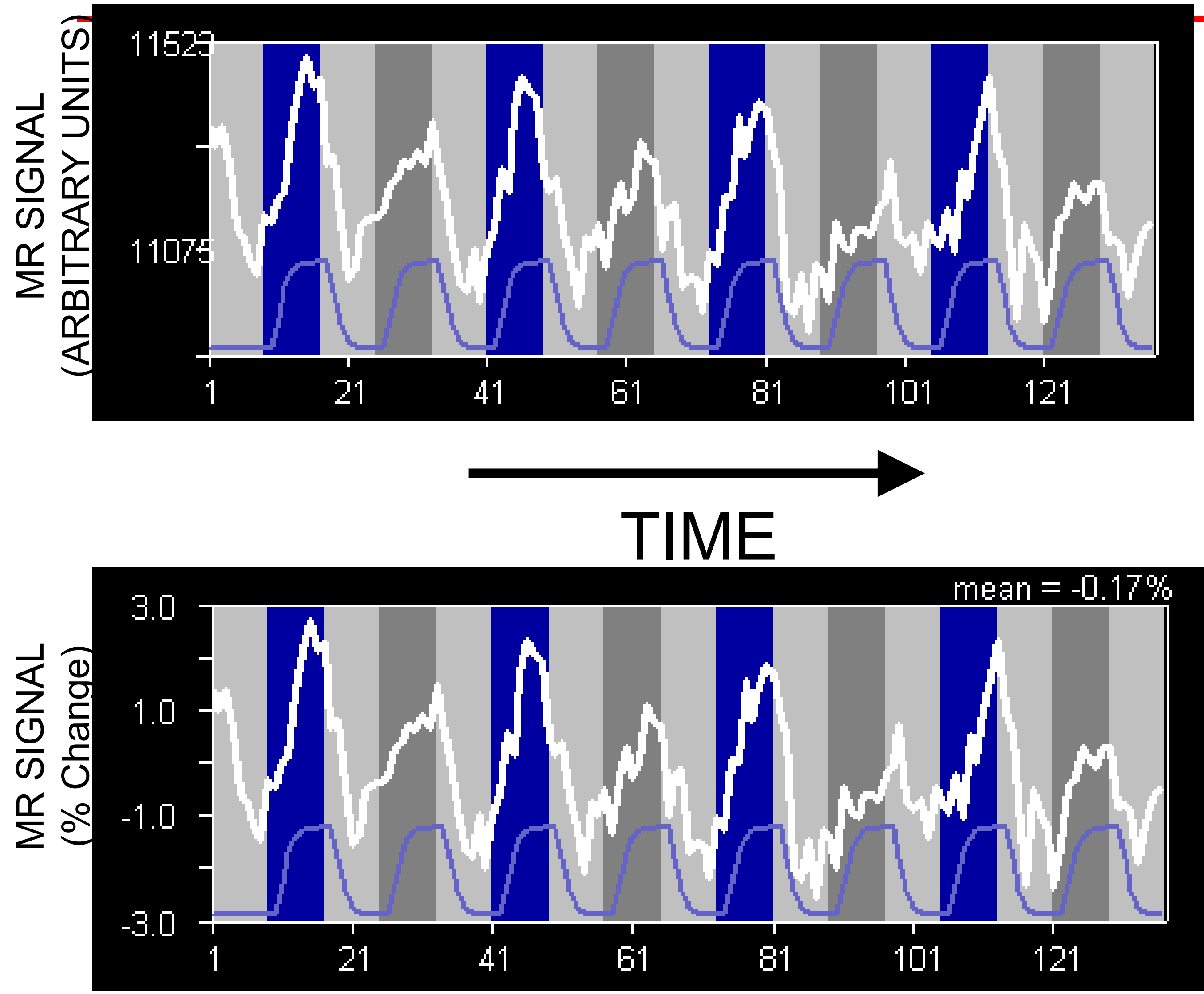
From: http://www.fmri4newbies.com/

Anatomic Slices Corresponding to Functional Slices



From: http://www.fmri4newbies.com/

Time Courses



Arbitrary signal varies from coil to coil, voxel to voxel, day to day, subject to subject

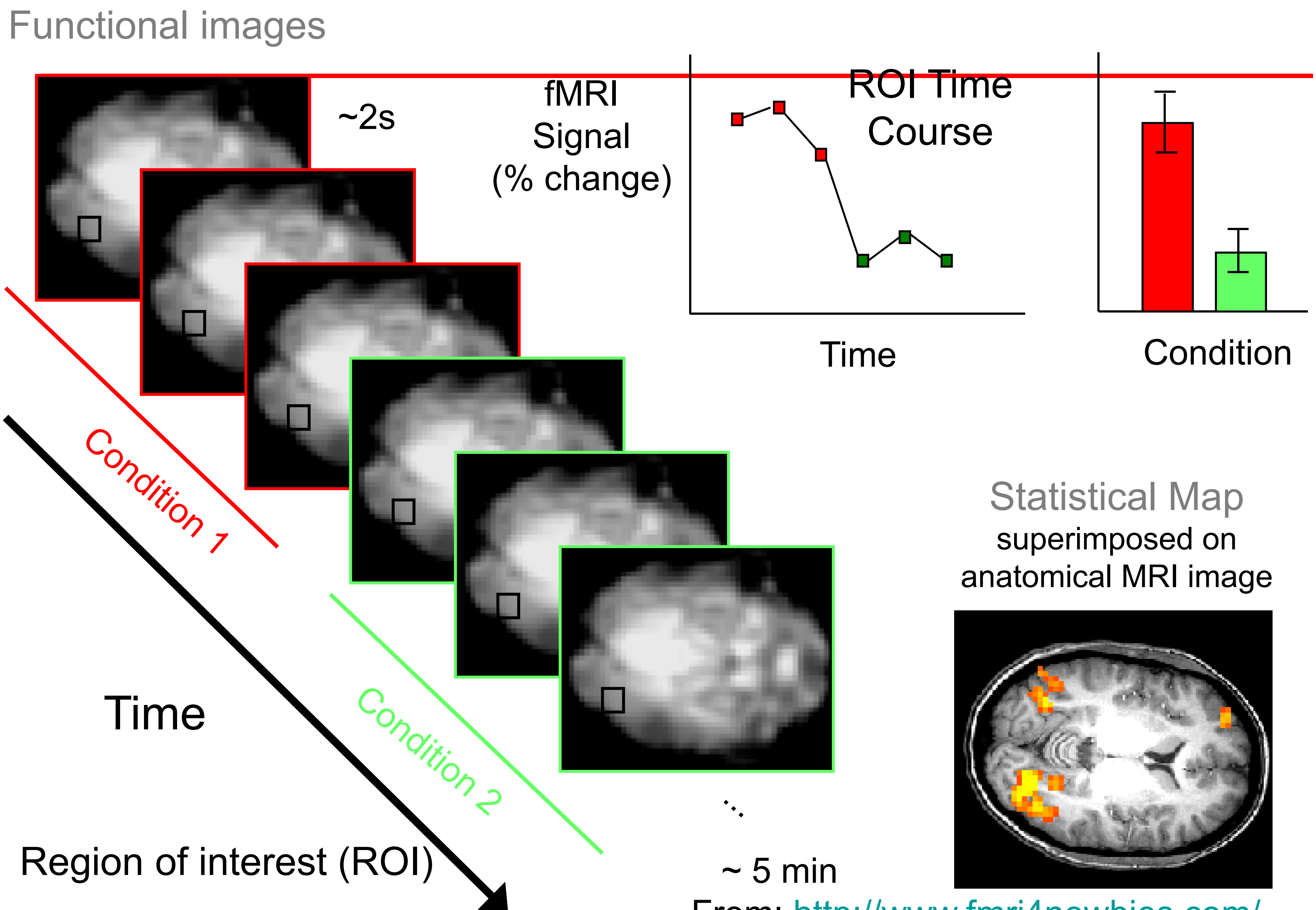
To make the y-axis more meaningful, we usually convert the signal into units of % change:

100*(x - baseline)/baseline

Changes are typically in the order of 0.5-4 %.

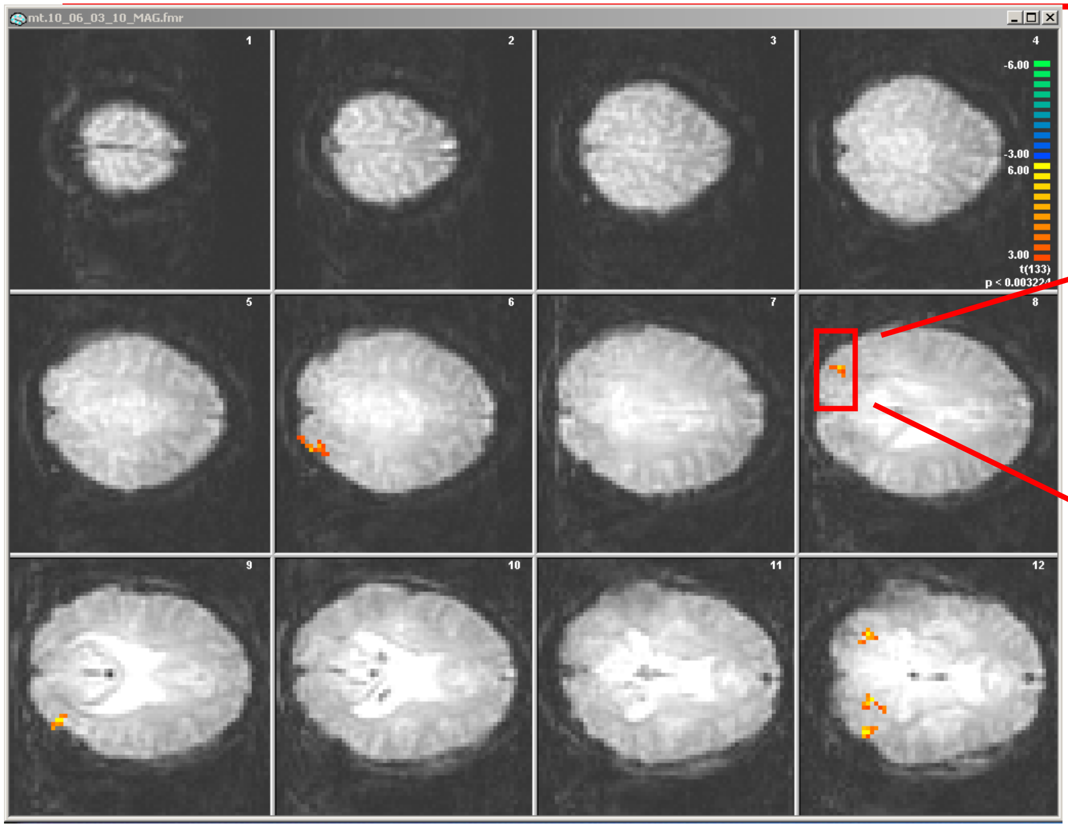
From: http://www.fmri4newbies.com/

Activation Statistics



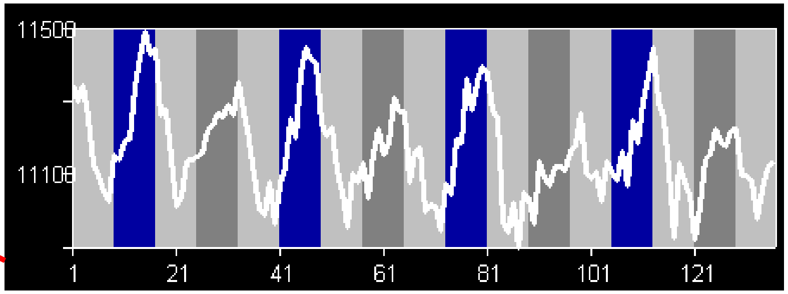
From: http://www.fmri4newbies.com/

Statistical Maps & Time Courses



Use stat maps to pick regions

Then extract the time course



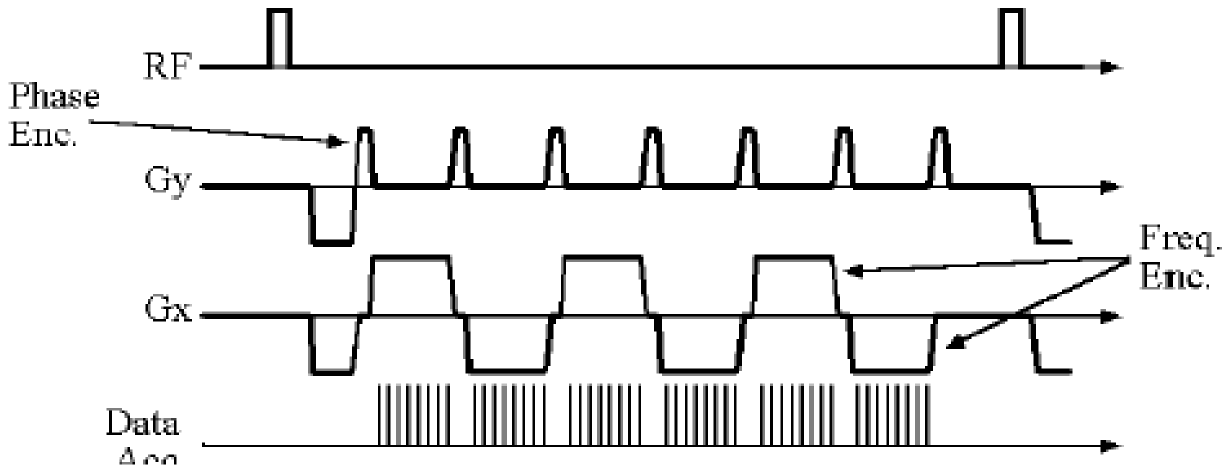
From: http://www.fmri4newbies.com/

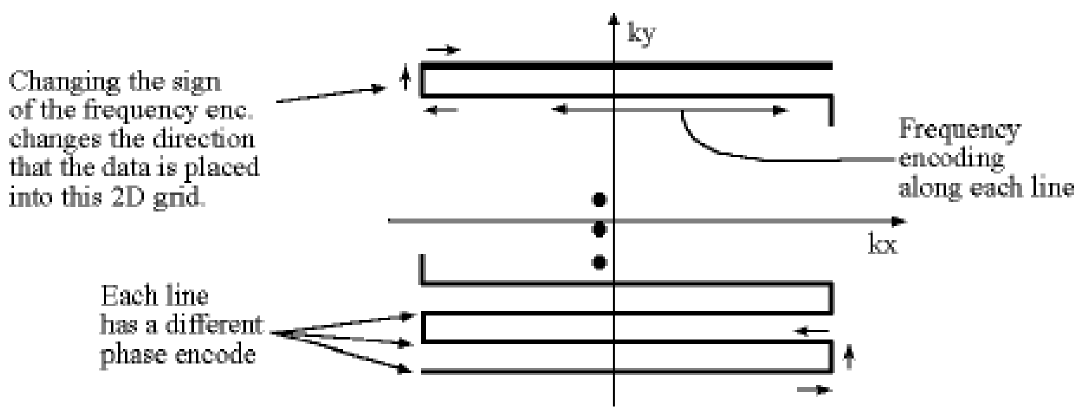
RF Sequences Used in fMRI

- Must image very fast
- Image the FID signal decrease due to T2*
- Typically use echo planar pulse sequence

Echo planar imaging

- Avoid going back to origin after each read-out
- “Single shot” imaging, popular in fMRI
- Spatial resolution limited by gradient switching time

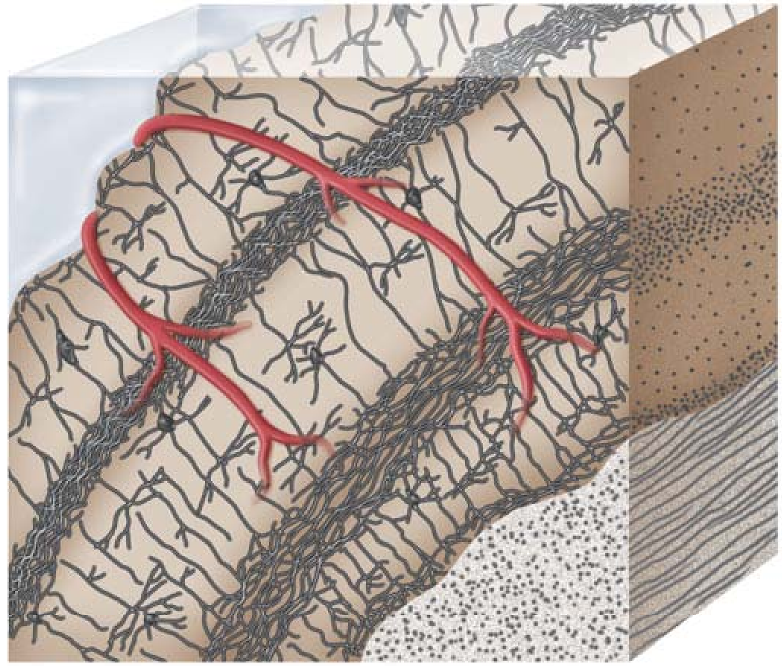




What Limits Spatial Resolution

- noise
  - smaller voxels have lower SNR
- head motion
  - the smaller your voxels, the more contamination head motion induces
- temporal resolution
  - the smaller your voxels, the longer it takes to acquire the same volume
    - 4 mm x 4 mm at 16 slices/sec
    - OR 1 mm x 1 mm at 1 slice/sec
- vasculature
  - depends on pulse sequences
    - e.g., spin echo sequences reduce contributions from large vessels
  - some preprocessing techniques may reduce contribution of large vessels (Menon, 2002, MRM)

Partial Voluming

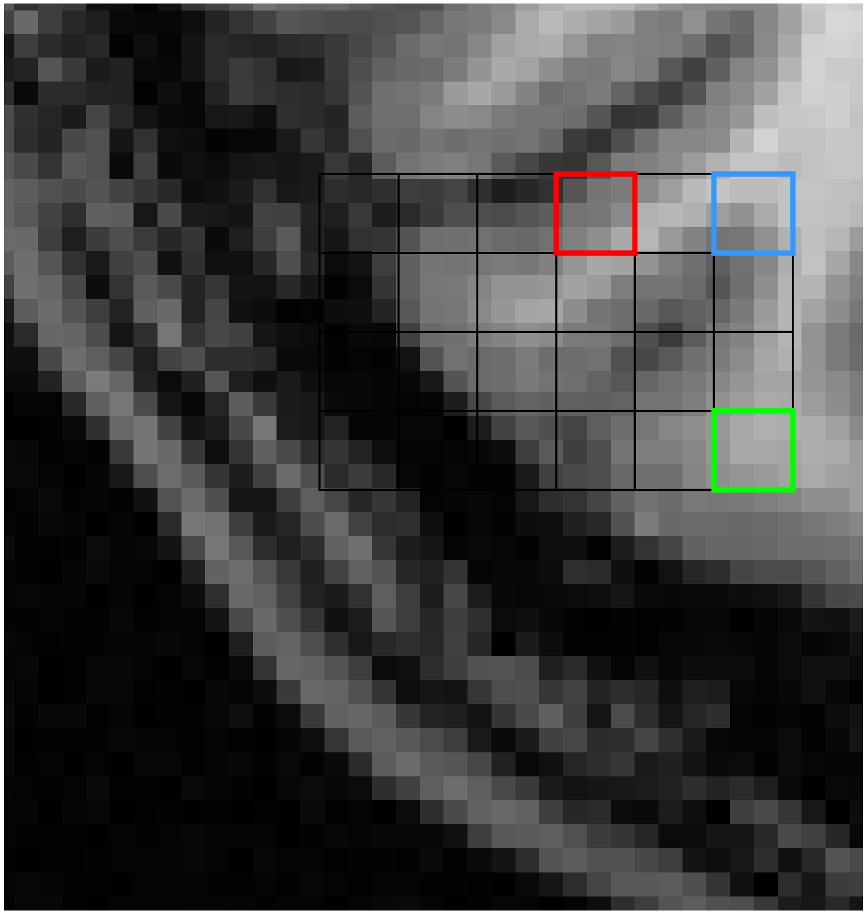


FUNCTIONAL MAGNETIC RESONANCE IMAGING, Figure 8.3 © 2004 Sinauer Associates, Inc.

- The fMRI signal occurs in gray matter (where the synapses and dendrites are)
- If your voxel includes white matter (where the axons are), fluid, or space outside the brain, you effectively water down your signal

Partial Voluming

Partial volume effects: The combination, within a single voxel, of signal contributions from two or more distinct tissue types or functional regions (Huettel, Song & McCarthy, 2004)



This voxel contains mostly gray matter

This voxel contains mostly white matter

This voxel contains both gray and white matter. Even if neurons within the voxel are strongly activated, the signal may be washed out by the absence of activation in white matter.

Partial voluming becomes more of a problem with larger voxel sizes

Worst case scenario: A 22 cm x 22 cm x 22 cm voxel would contain the whole brain

Acknowledgments

- I am extremely grateful to:
  - Prof. Pippa Storey (NYU, Radiology) who provided some of the slides about image quality
  - Dr. Pablo Velasco (NYU, Center for Brain Imaging) and Prof. Yao Wang (NYU-Poly) who provided some of the slides about fMRI
  - Prof. Eric Sigmund (NYU, Radiology) who provided some of the slides about diffusion MRI

Homework

- Reading:
  - Prince and Links, Medical Imaging Signals and Systems, Review Chap. 13
  - Note down all the corrections for Ch. 13 on your copy of the textbook based on the provided errata (see Course website or book website for update)
- Problems
  - No Problems

See you at the end of the course!