Medical Imaging Instrumentation & Image Analysis – MRI subportion

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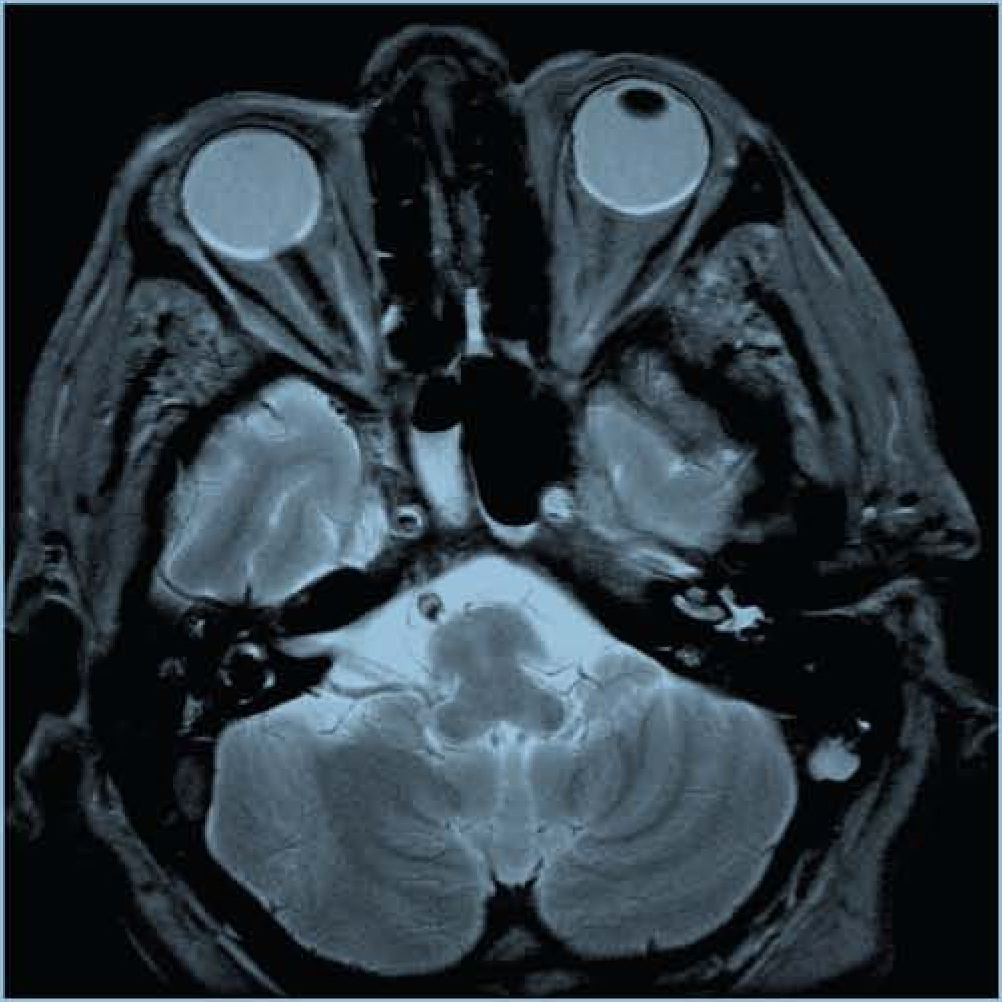
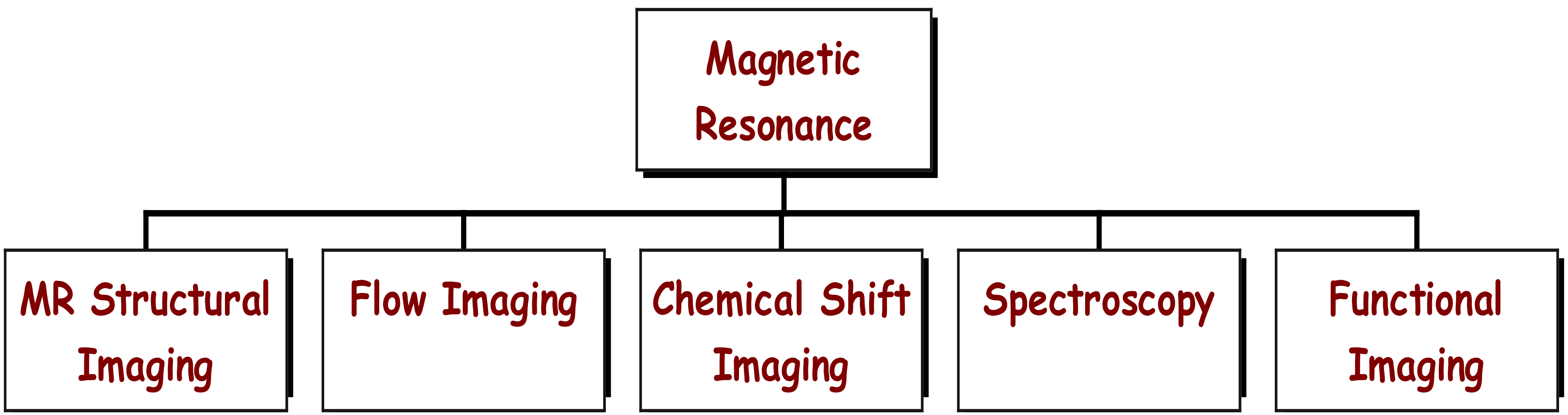
Basics of Magnetic Resonance Imaging



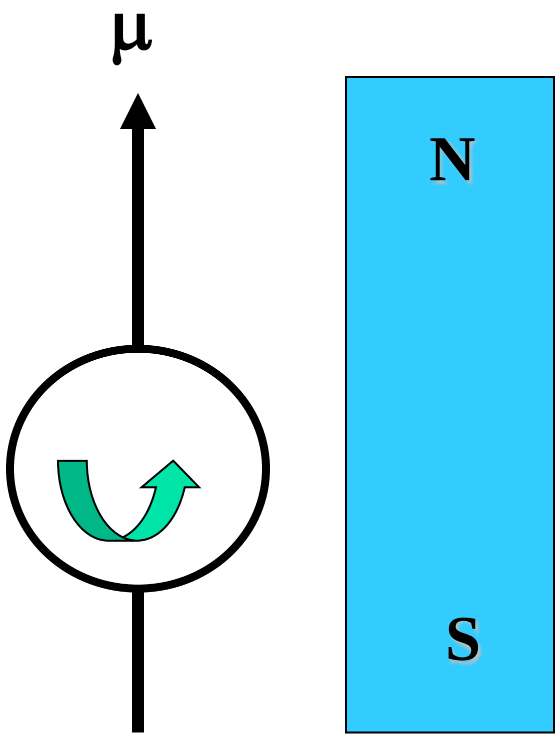
Mark Cohen
http://www.brainmapping.org/Ma

Magnetic Resonance Imaging

Basic Principle: The electromagnetic induction based rf signals are collected through nuclear magnetic resonance from the excited nuclei with magnetic moment and angular momentum present in the body. Most common is proton density imaging.



Spinning Protons



Protons with a spinning property behave like small magnets.

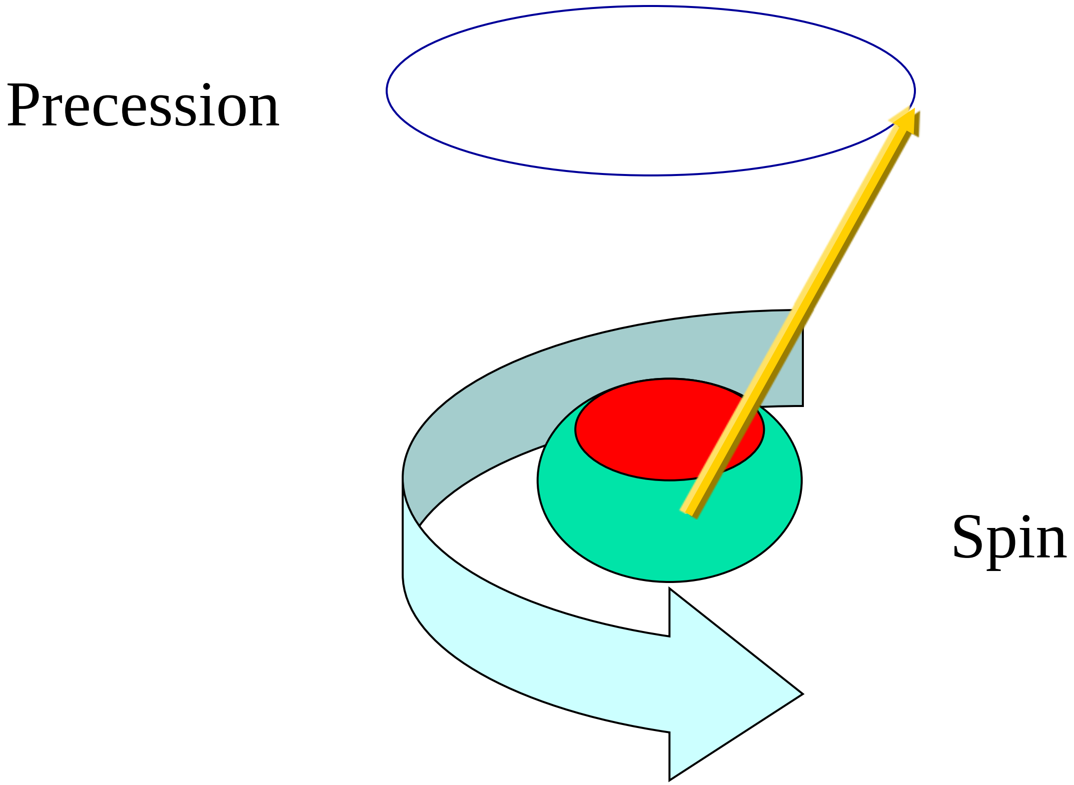
Spinning around their own axes results in generation of a magnetic moment, μ.

When placed in external magnetic field, spinning protons align themselves either along or against the external magnetic field.

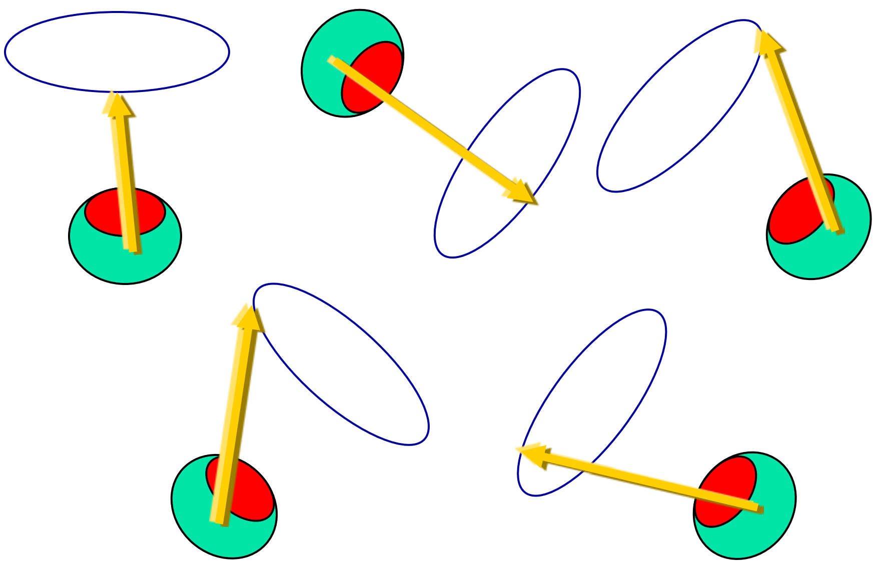
In addition, placing spinning proton in an external magnetic field causes the magnetic moment to precess around an axis parallel to the field direction.

Spinning Proton

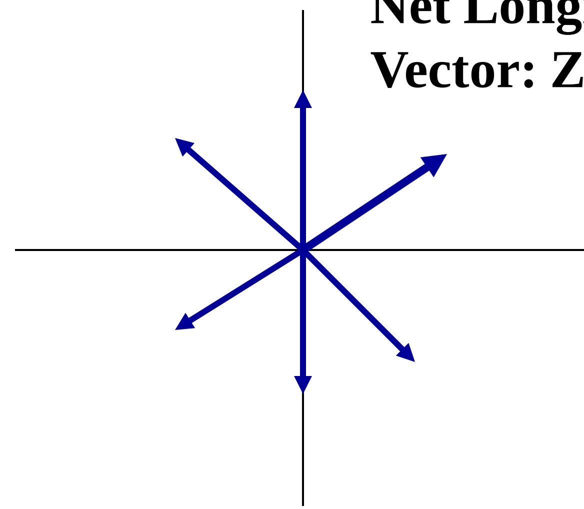
Protons possessing properties of angular and magnetic moments provide signals for nuclear magnetic resonance

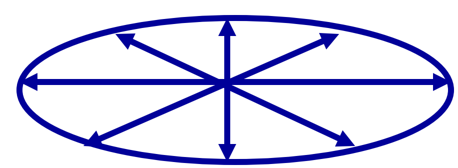


Protons With Random Effect



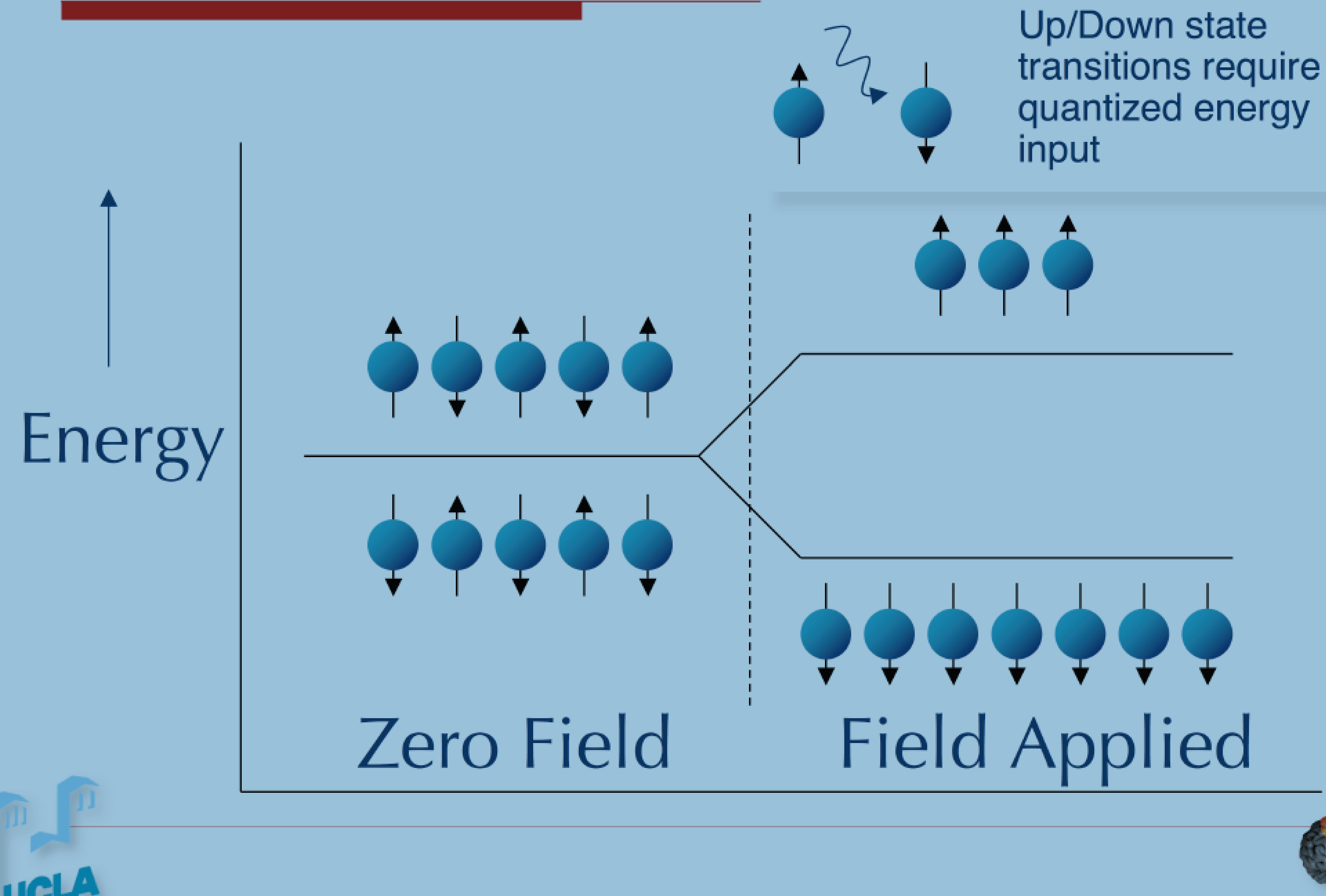
Net Longitudinal Vector: Zero





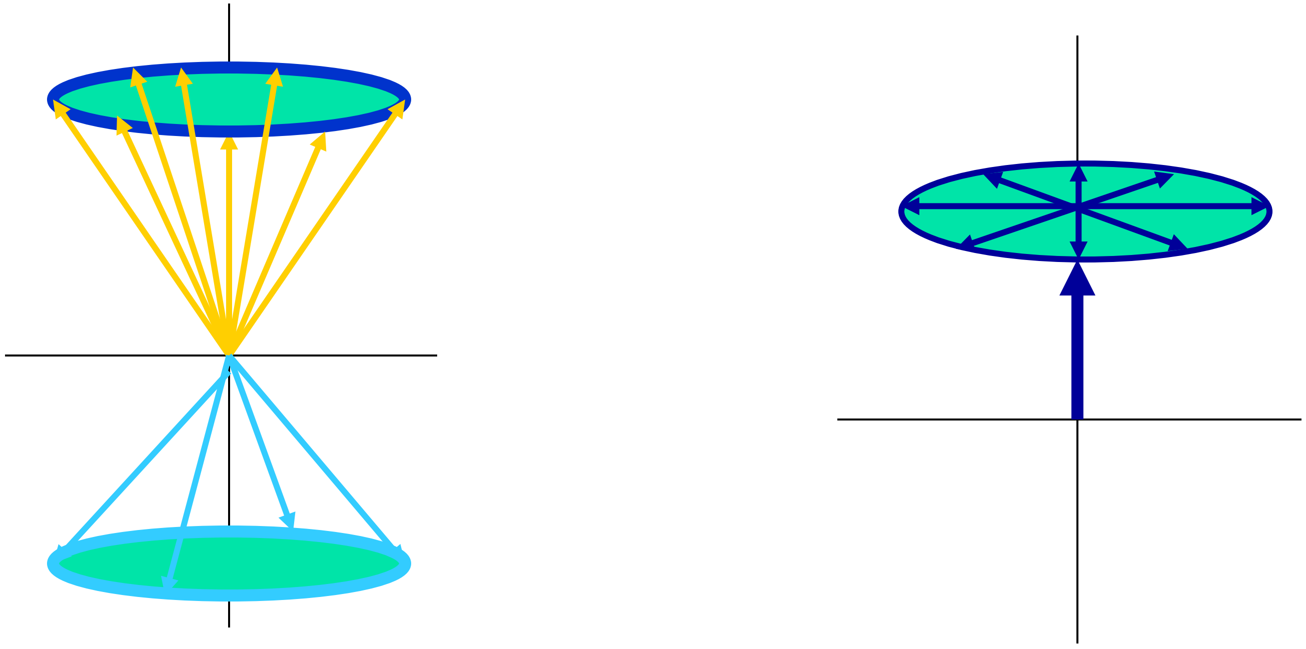
Net Transverse Vector: Zero

Transition to Equilibrium

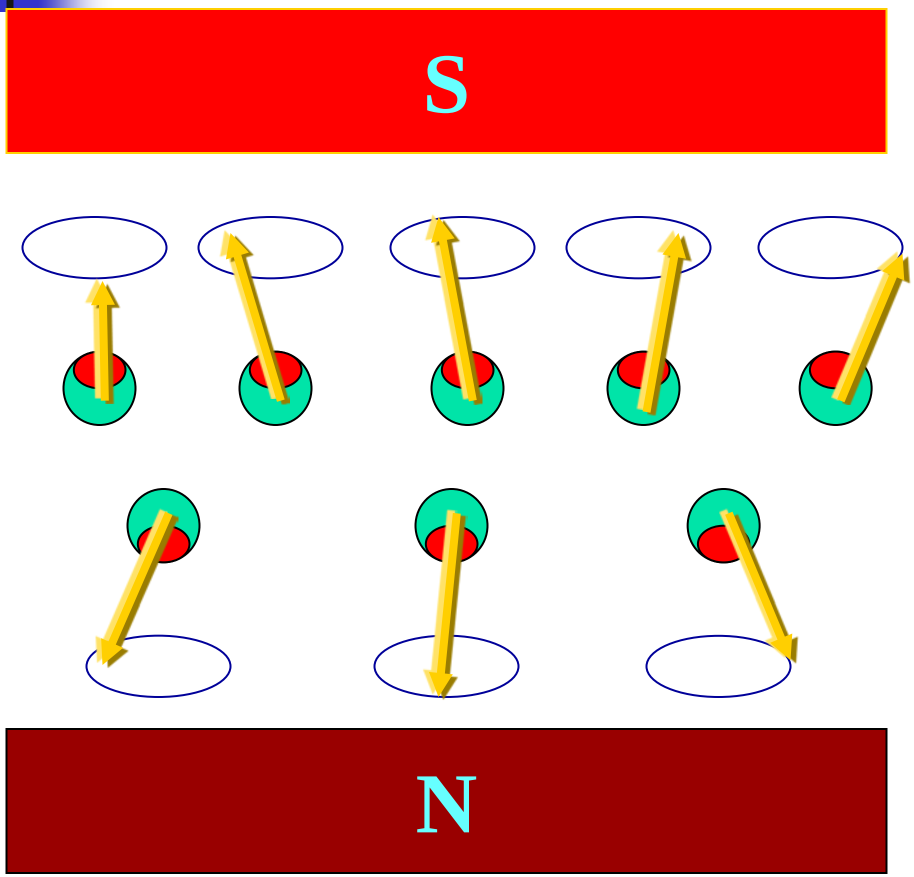


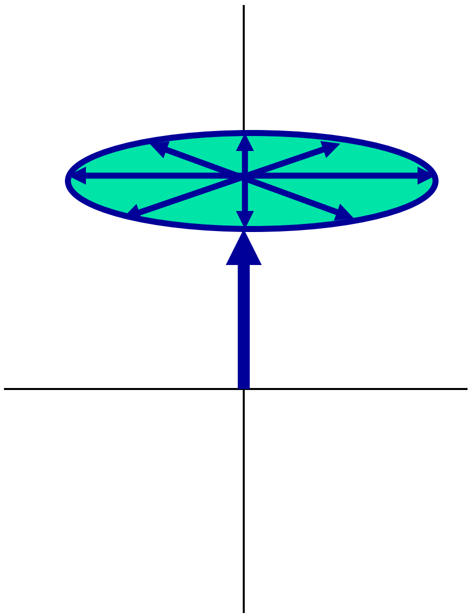
Net Vector Under Thermal Equilibrium

Larmor (Precession) Frequency $\omega = \gamma \mathbf{H}$

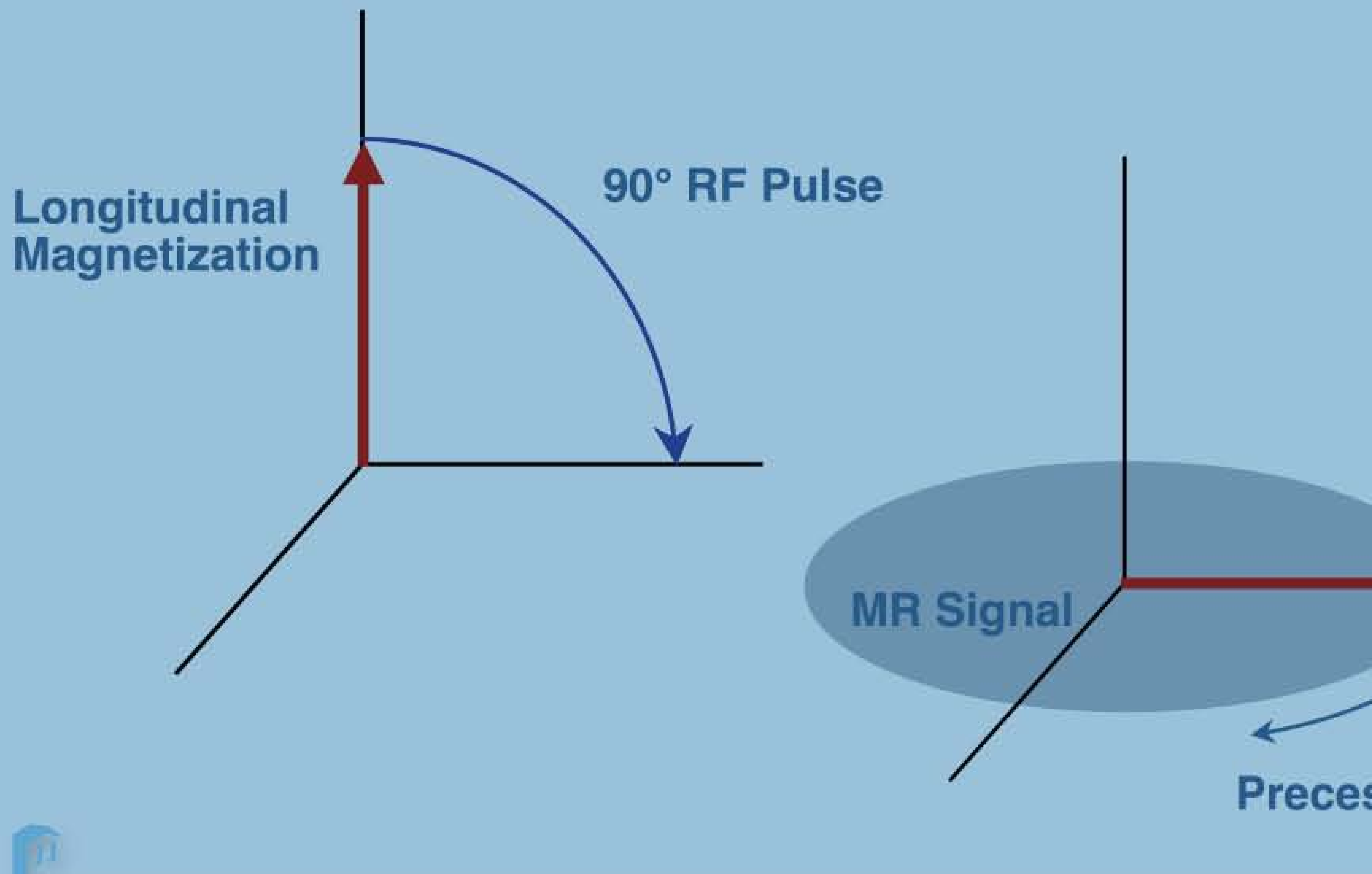


Protons Under Thermal Equilibrium

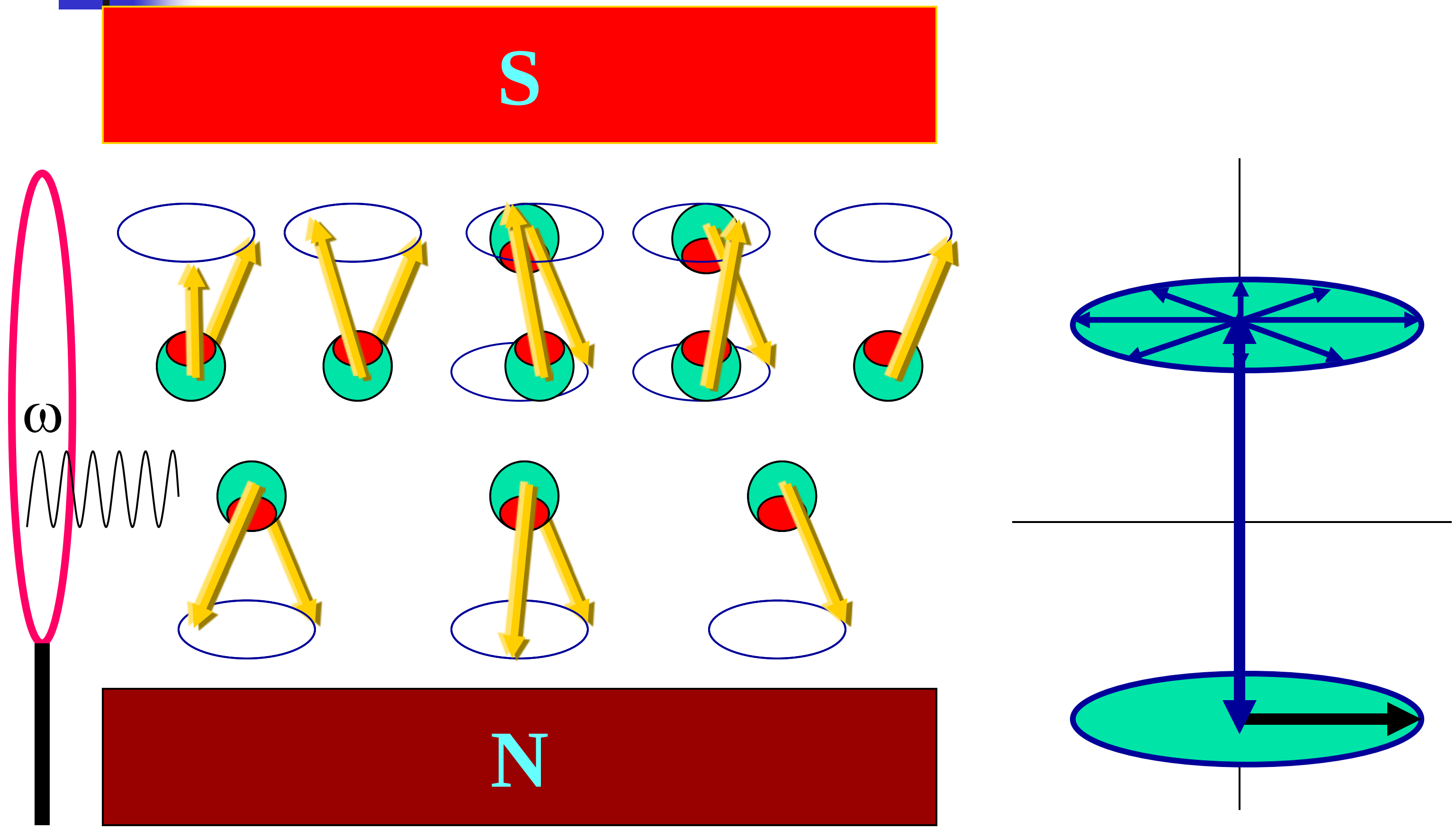




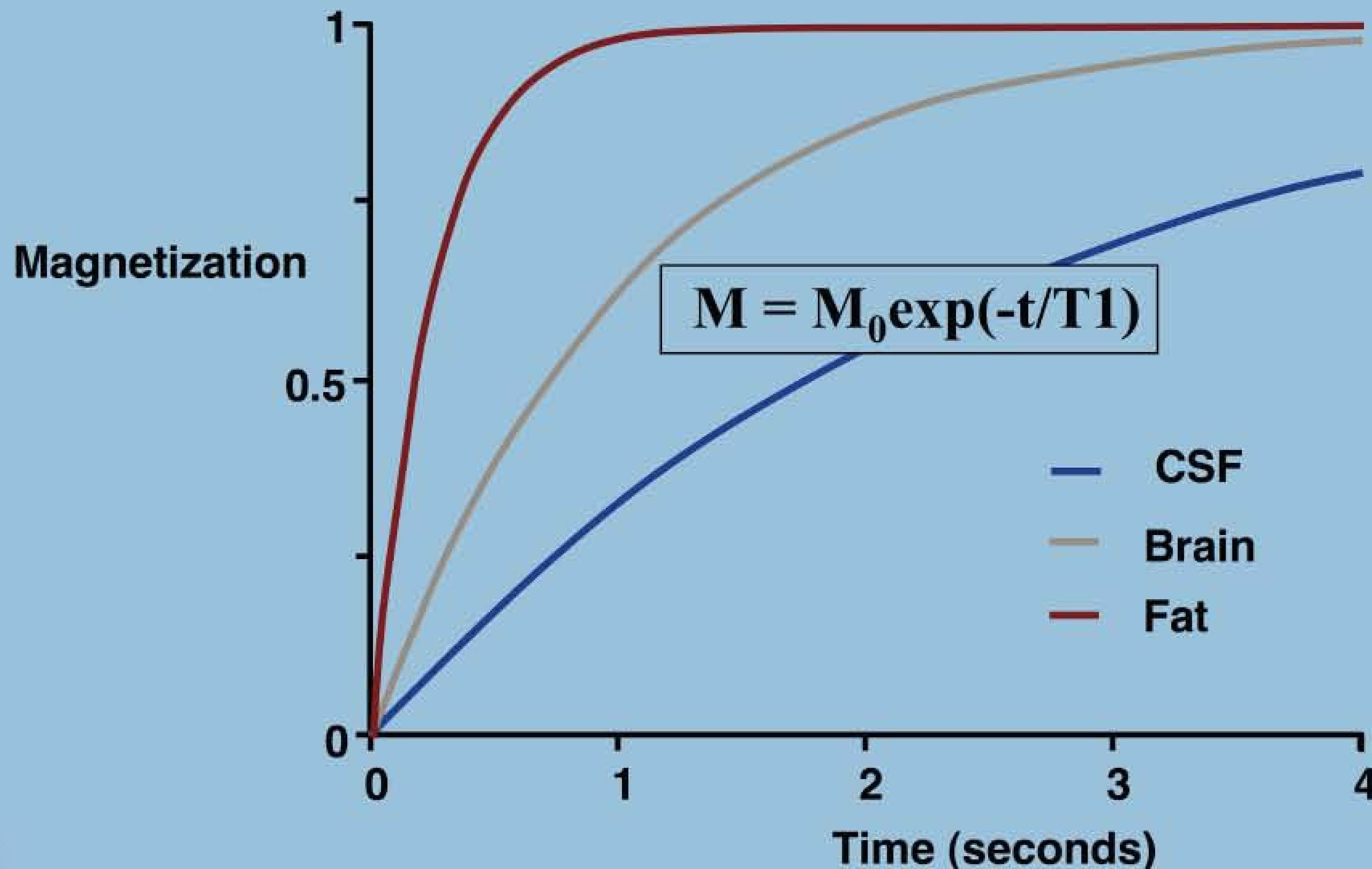
An RF Pulse Converts Longitudinal Magnetization to Signal



Protons With External RF Excitation: 180 Degree Pulse

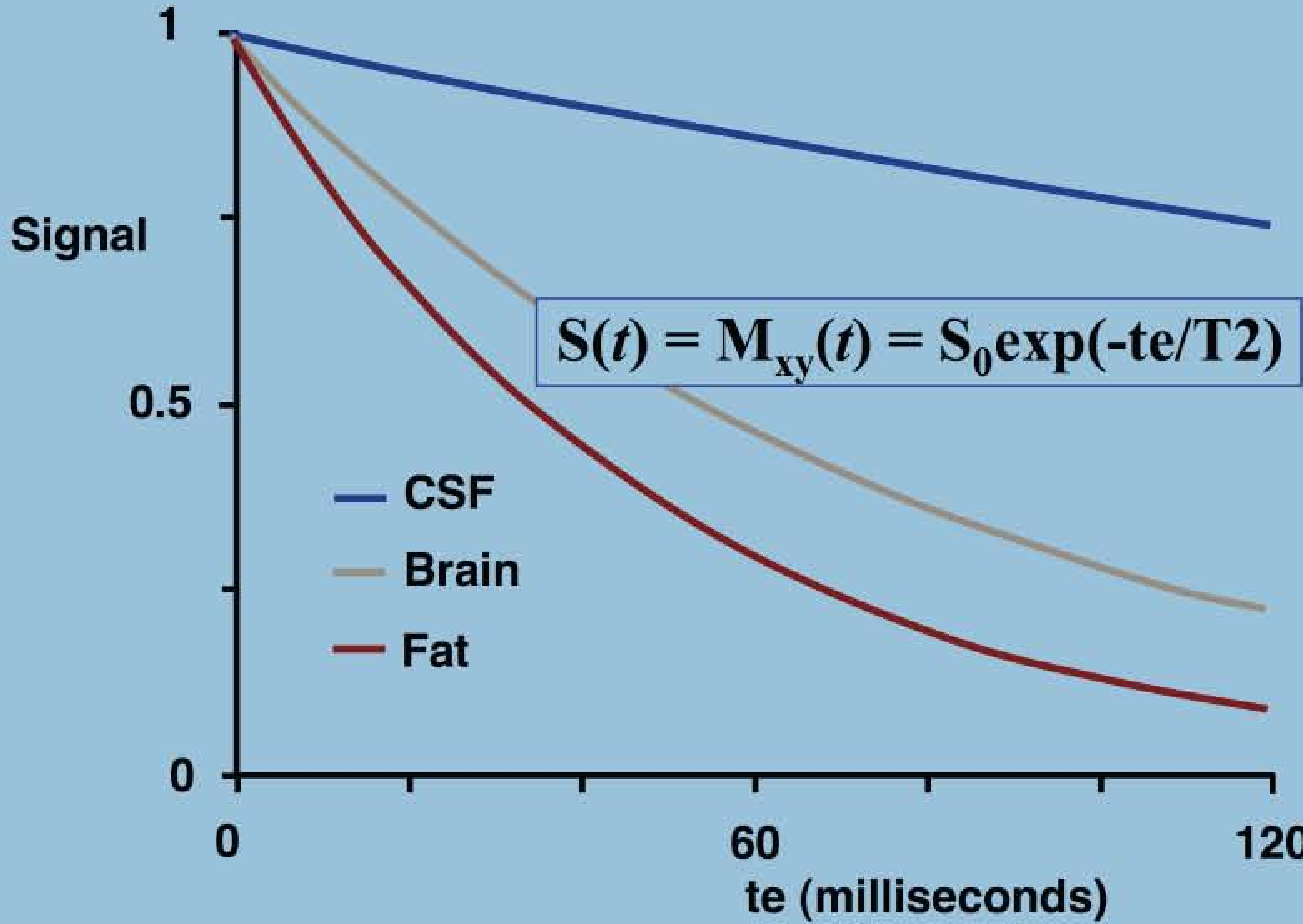


T1

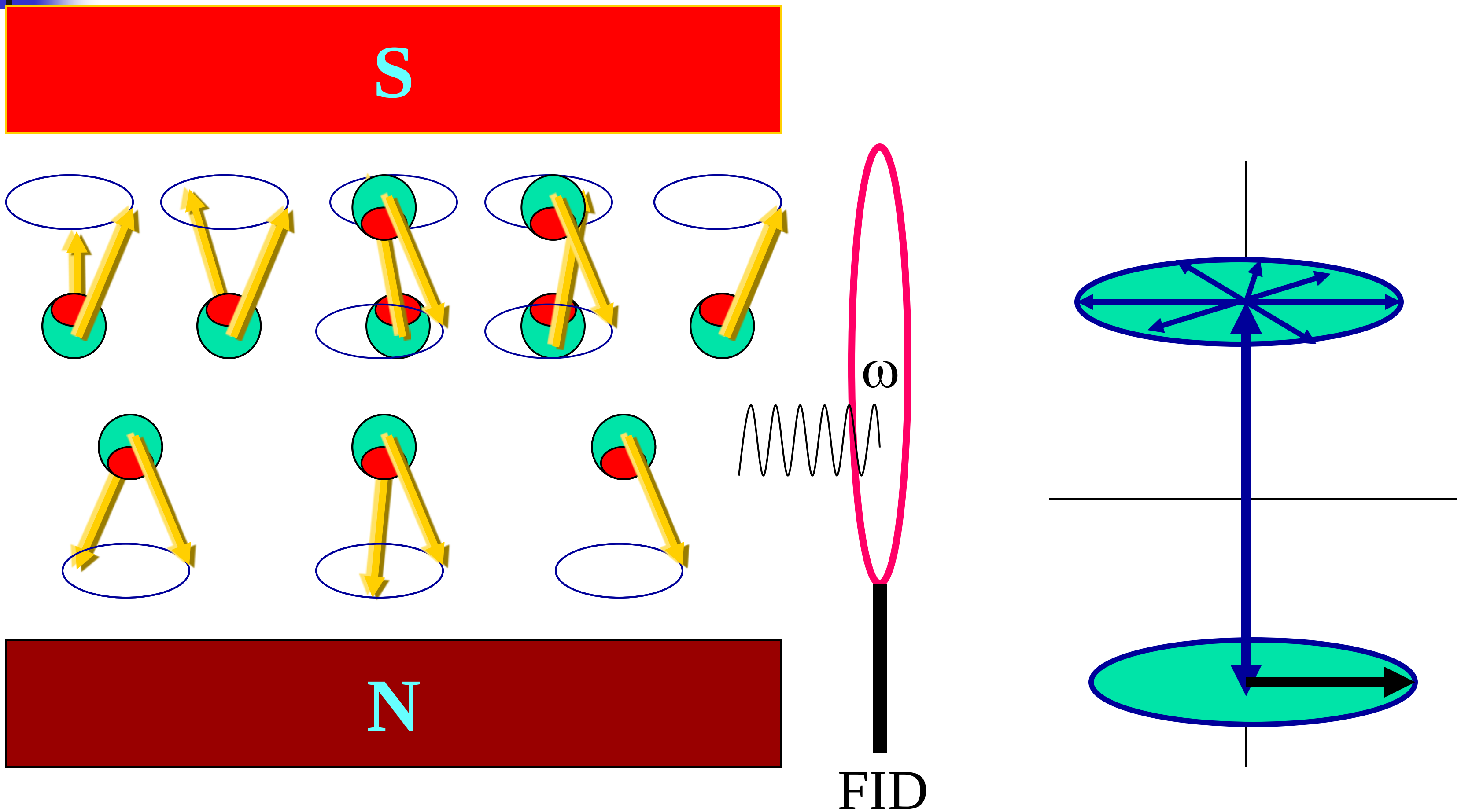


1
0.5
0
Magnetization
M = M0exp(-t/T1)
CSF
Brain
Fat
0
1
2
3
4
Time (seconds)

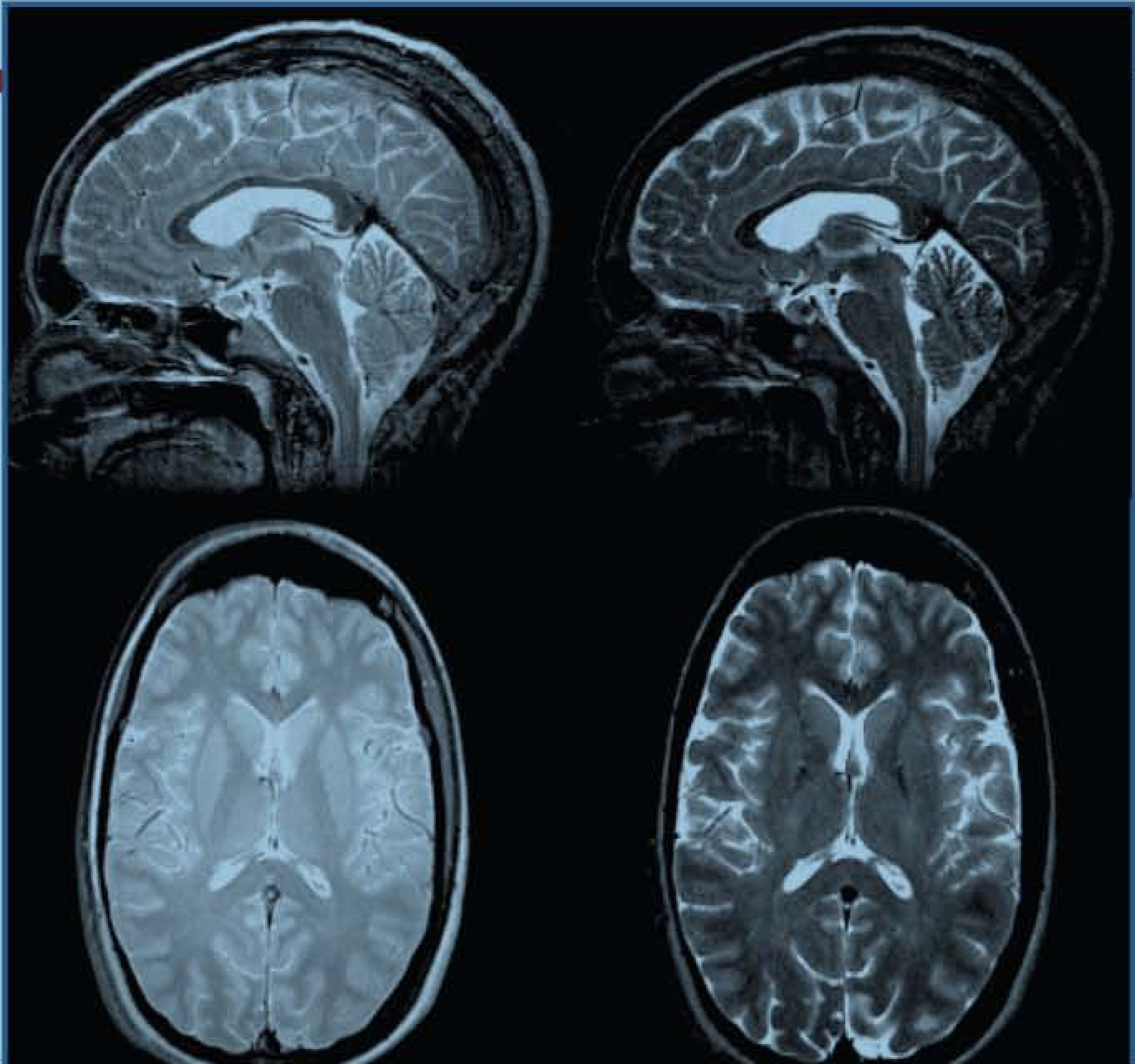
T2 and TE


1
Signal
0.5
0
CSF
Brain
Fat
S(t) = M_xy(t) = S_0exp(-te/T2)
0
60
120
te (milliseconds)

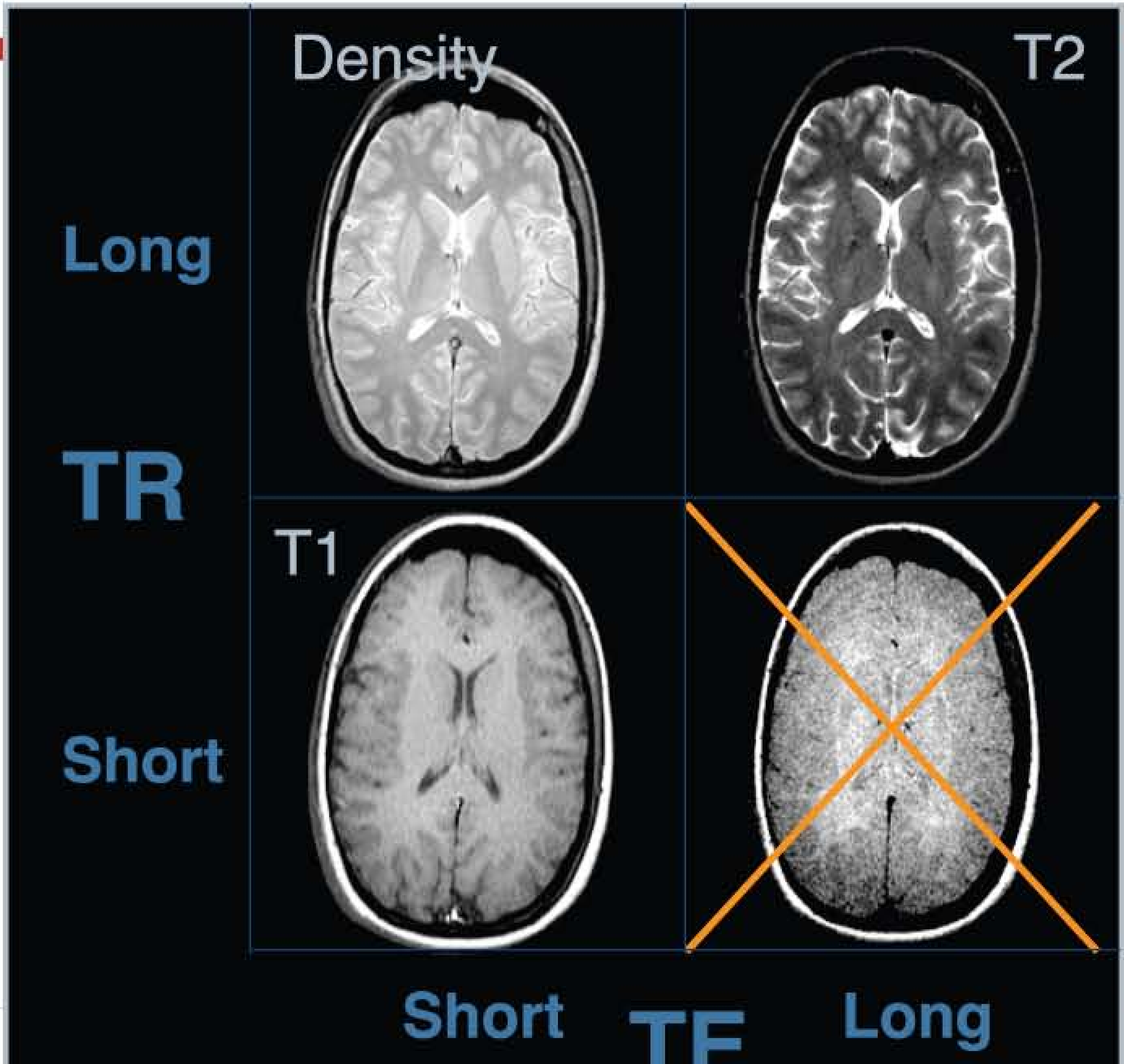
Relaxation Process Provides FID or MR Signal



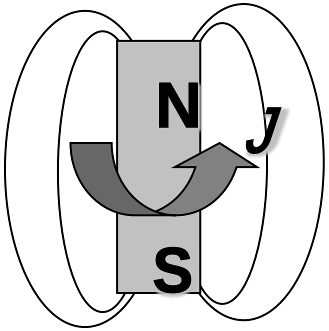
Effects of TE at long TR

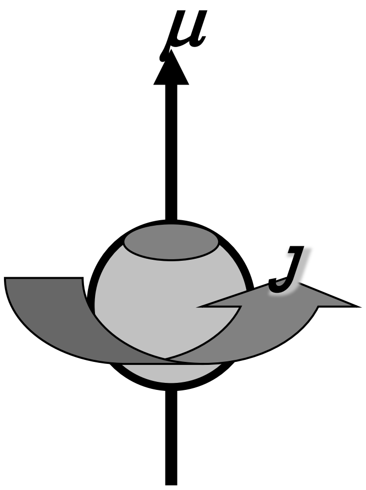


Contrast, TR and TE



Magnetic moment





- The spin angular moment, *J*, and the magnetic moment, μ, is described by

$$\vec{\mu} = \gamma \vec{J}$$

where γ is a gyromagnetic ratio defined in MHz/T.

For hydrogen proton

γ=42.58 MHz/T

Angular Momentum

The torque generated by the interaction of magnetic moment of a proton and the external magnetic field is equal to the rate of change of angular momentum and can be given by the equation of motion for isolated spin as

$$\frac{d\vec{J}}{dt} = \mu \times \vec{H}_0 = \mu \times H_0 \vec{k}$$

$$\vec{\mu} = \gamma \vec{J}$$

$$\frac{d\overrightarrow{\mu}}{dt} = \gamma \, \mu \times H_0 \vec{k}$$

Total Magnetic Moment

Assuming N to be the total number of spinning nuclei in the object being imaged, a stationary magnetization vector, $\vec{M}$

can be defined from the available magnetic moments as

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

$$\vec{M} = M_x \vec{i} + M_y \vec{j} + M_z \vec{k}$$

and

$$\vec{M}_r = M_{x'} \vec{i}\,' + M_{y'} \vec{j}\,' + M_{z'} \vec{k}\,'$$

Rotating Field

$$\vec{i}' = \cos(\omega t)\vec{i} - \sin(\omega t)\vec{j}$$

$$\vec{j}' = \sin(\omega t)\vec{i} + \cos(\omega t)\vec{j}$$

$$\vec{k}' = \vec{k}$$

$$\begin{bmatrix} M_{x'} \\ M_{y'} \\ M_{z'} \end{bmatrix} = \begin{bmatrix} \cos\ \omega t & -\sin\ \omega t & 0 \\ \sin\ \omega t & \cos\ \omega t & 0 \\ 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} M_x \\ M_y \\ M_z \end{bmatrix}$$

The transverse magnetization vector in the rotating frame can be written as

$$M_{x',y'} = M_{x,y} e^{i\omega t}$$

where

$$M_{x,y} = M_x + iM_y \quad \text{and} \quad M_{x',y'} = M_{x'} + iM_{y'}$$

Oscillating RF Field

Let us assume that *H1r* and *H1* are, respectively, the RF field in the rotating frame and the stationary coordinates systems.

An oscillating RF field causing nuclear excitation can be expressed as

$$H_{1r}(t) = H_1(t)e^{i\omega t}$$

where

$$H_1 = H_{1,x} + iH_{1,y} \text{ and } H_{1r} = H_{1,x'} + iH_{1,y'}$$

Bloch Equation

The relationship between the rates of change of stationary magnetization vector $\vec{M}$ and rotating magnetization vector $\vec{M}_r$ can then be expressed as

$$\frac{d\vec{M}}{dt} = \frac{\partial \vec{M}_r}{\partial t} + \omega \times \vec{M}_r$$

During the RF pulse (nuclear excitation phase), the rate of change in the net stationary magnetization vector can be expressed as (the Bloch Equation):

$$\frac{d\vec{M}}{dt} = \gamma \, \vec{M} \times \vec{H}$$

Total Response

Considering the total response of the spin system in the presence of an external magnetic field along with the RF pulse for nuclear excitation followed by the nuclear relaxation phase, the change of the net magnetization vector can be expressed as

$$\frac{d\vec{M}}{dt} = \gamma \vec{M} \times \vec{H} - \frac{M_x \vec{i} - M_y \vec{j}}{T_2} - \frac{(M_z - M_z^0)\vec{k}}{T_1}$$

$\vec{M}_z^0$ is the net magnetization vector in thermal equilibrium in the presence of an external magnetic field H_0 only, and T_1 and T_2 are, respectively, the longitudinal (spin-lattice) and transverse (spin-spin) relaxation times in the nuclear relaxation phase when excited nuclei return to their thermal equilibrium state.

Relaxation Process

$$M_{x,y}(t) = M_{x,y}(0)e^{-t/T_2}e^{-i\omega_0 t}$$

$$M_z(t) = M_z^0(1 - e^{-t/T_1}) + M_z(0)e^{-t/T_1}$$

$$\text{where } M_{x,y}(0) = M_{x',y'}(0)e^{-i\omega_0 \tau_p}$$

$M_{x,y}(0)$ represents the initial transverse magnetization vector with the time set to zero at the end of the RF pulse of duration tp.

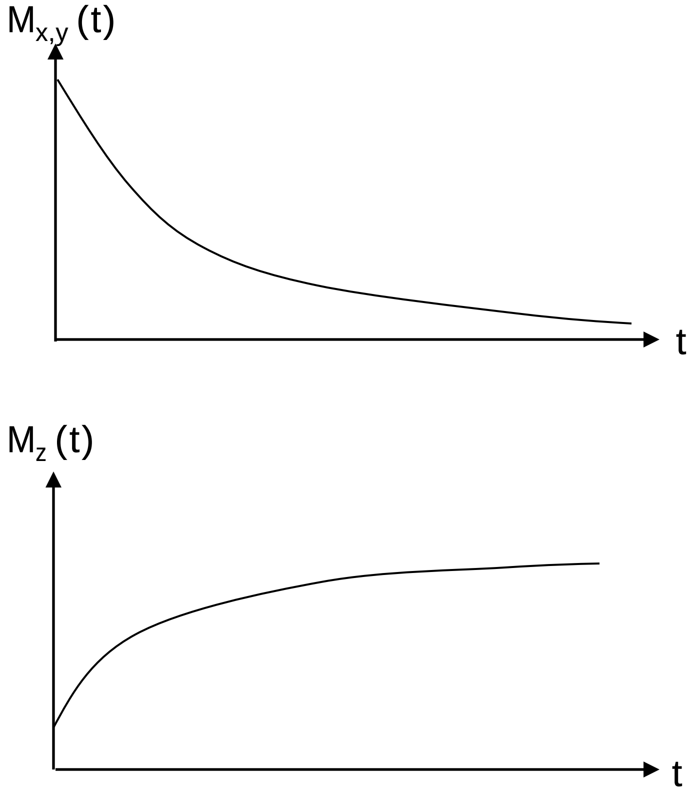
Signal Induction

$$\phi(t) = \int_{object} \vec{H}_r(\mathbf{r}) \cdot \vec{M}(\mathbf{r},t)d\mathbf{r}$$

$$\mathbf{r} = x\vec{i} + y\vec{j} + z\vec{k}$$

$$V(t) = -\frac{\partial \phi(t)}{\partial t} = -\frac{\partial}{\partial t}\int_{object} \vec{H}_r(\mathbf{r}) \cdot \vec{M}(\mathbf{r},t)d\mathbf{r}$$

Relaxation Vectors



Relaxation Times

Tissue	T1 msec	T2 msec	SD %
Fat	150	150	10.9
Liver	250	44	10.0
White Matter	300	133	11.0
Gray Matter	475	118	10.5
Blood	525	261	10.0
CSF	2000	250	10.8

Signal Coding

$$\mathbf{G}(t) = G_x(t)\vec{i} + G_y(t)\vec{j} + G_z(t)\vec{k}$$

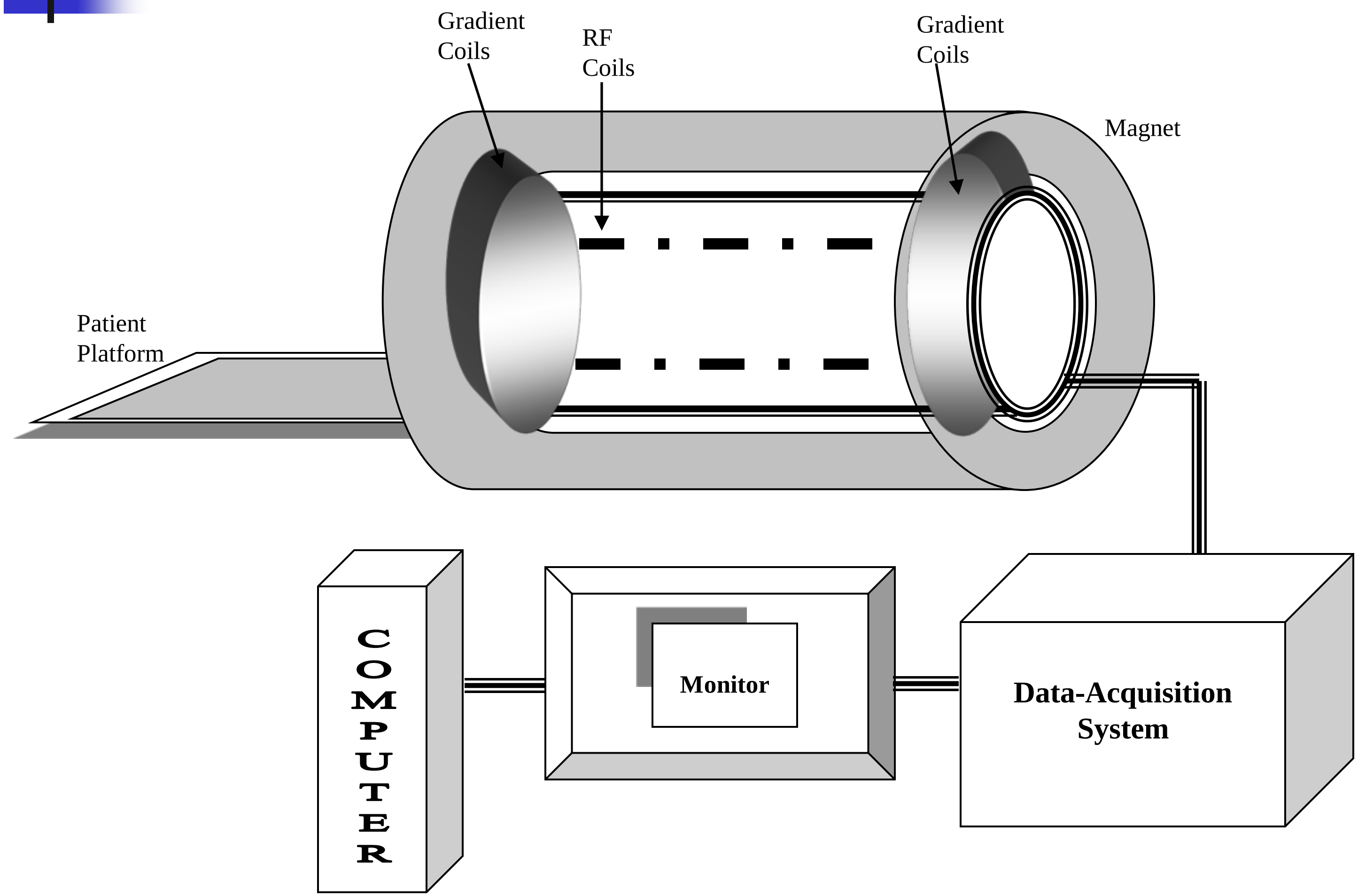
$$S(t) = \int \vec{M}(\mathbf{r},t)d^3r$$

$$\vec{M}(\mathbf{r},t) = \vec{M}_0\rho(\mathbf{r})e^{-i\gamma\mathbf{r}\cdot\int_0^t \mathbf{G}(t')dt'}$$

$$S(\omega_x,\omega_y,\omega_z) = \vec{M}_0\iiint\rho(x,y,z)e^{-i(\omega_x x+\omega_y y+\omega_z z)}dxdydz$$

$$\rho(x,y,z) = \vec{M}_0\iiint S(\omega_x,\omega_y,\omega_z)e^{i(\omega_x x+\omega_y y+\omega_z z)}d\omega_x d\omega_y d\omega_z$$

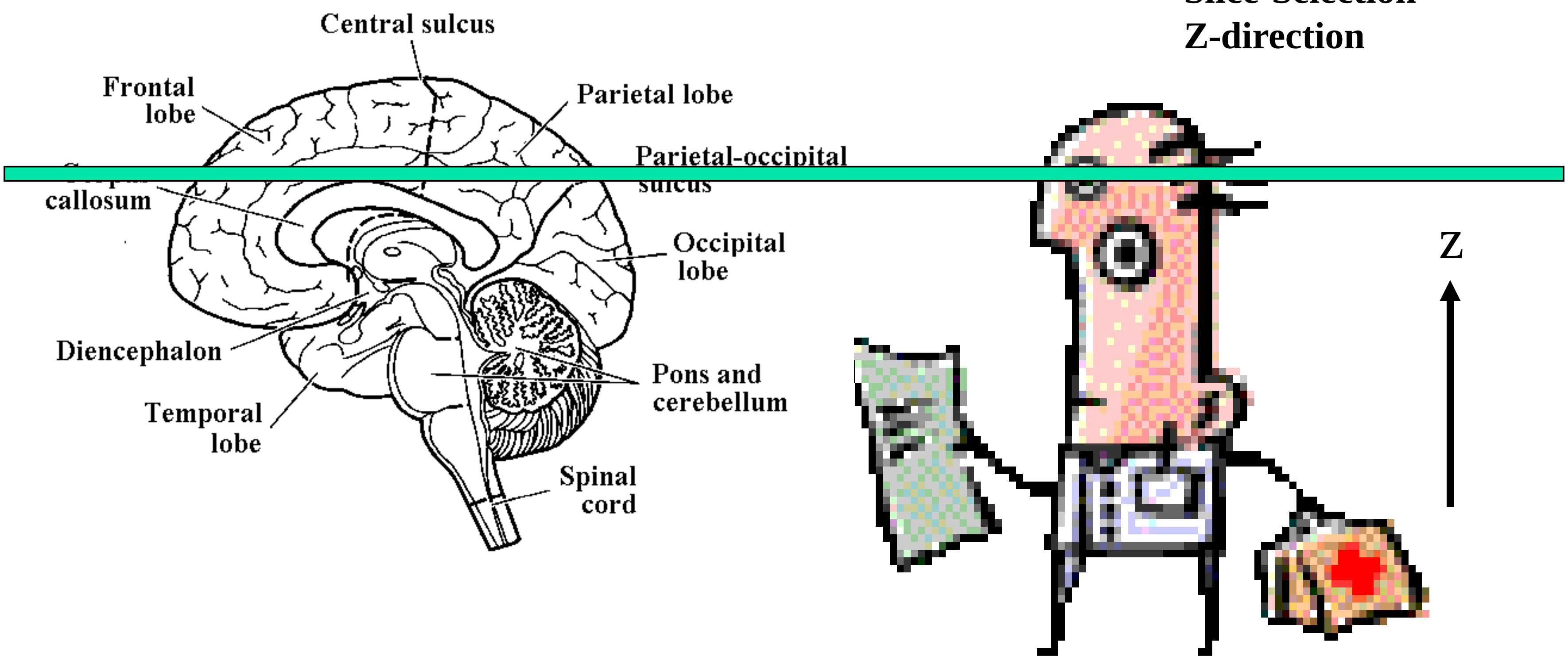
MR Imaging


Gradient
Coils
RF
Coils
Gradient
Coils
Magnet
Patient
Platform
COMPUTER
Monitor
Data-Acquisition
System

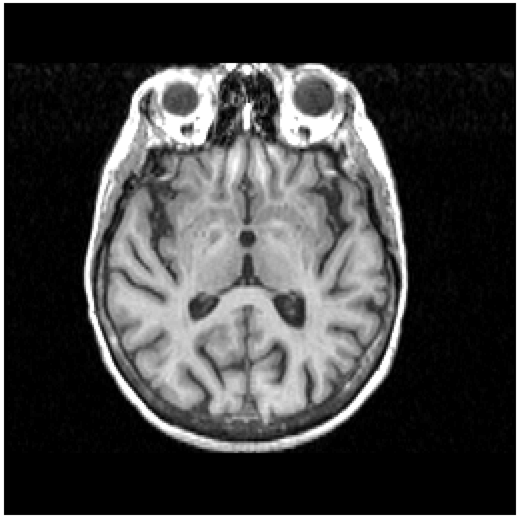

new balance
new balance
SIGNA

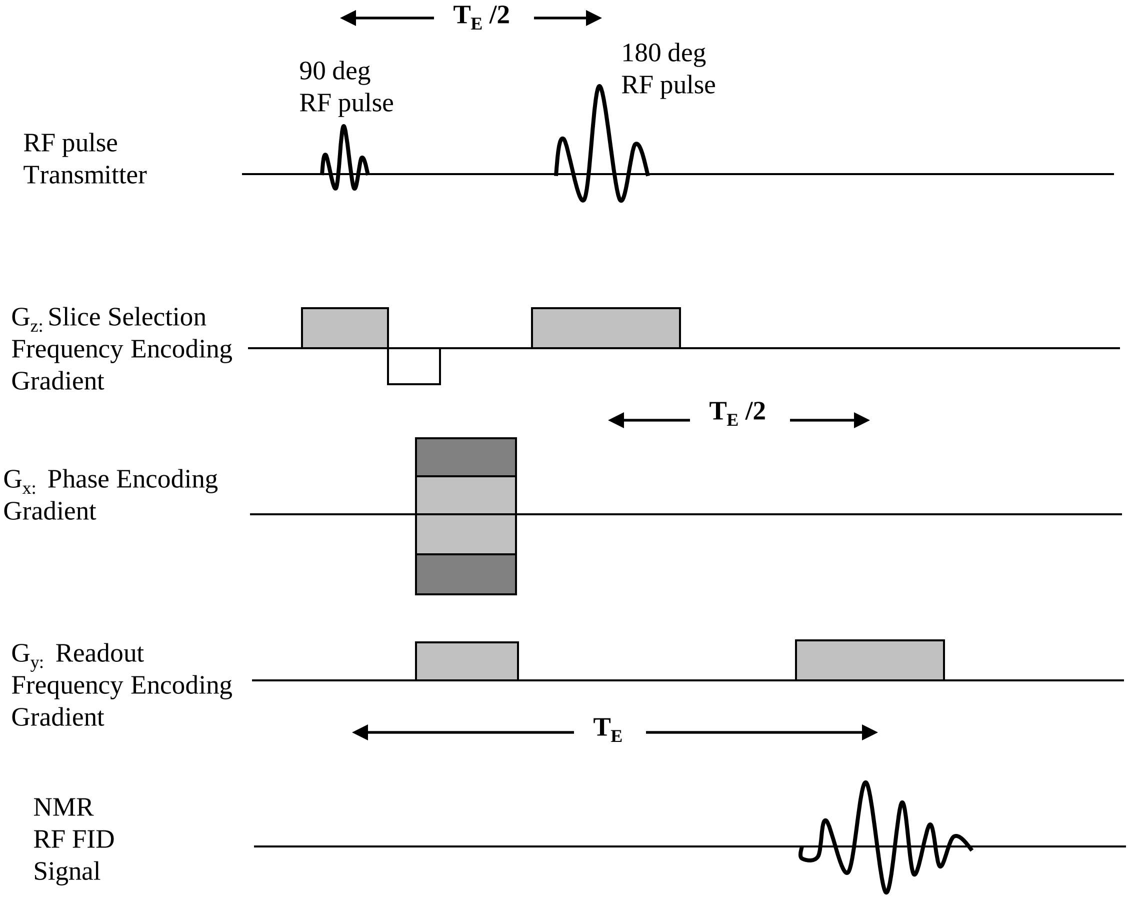
Spin-Echo Imaging Sequence

Slice-Selection
Z-direction

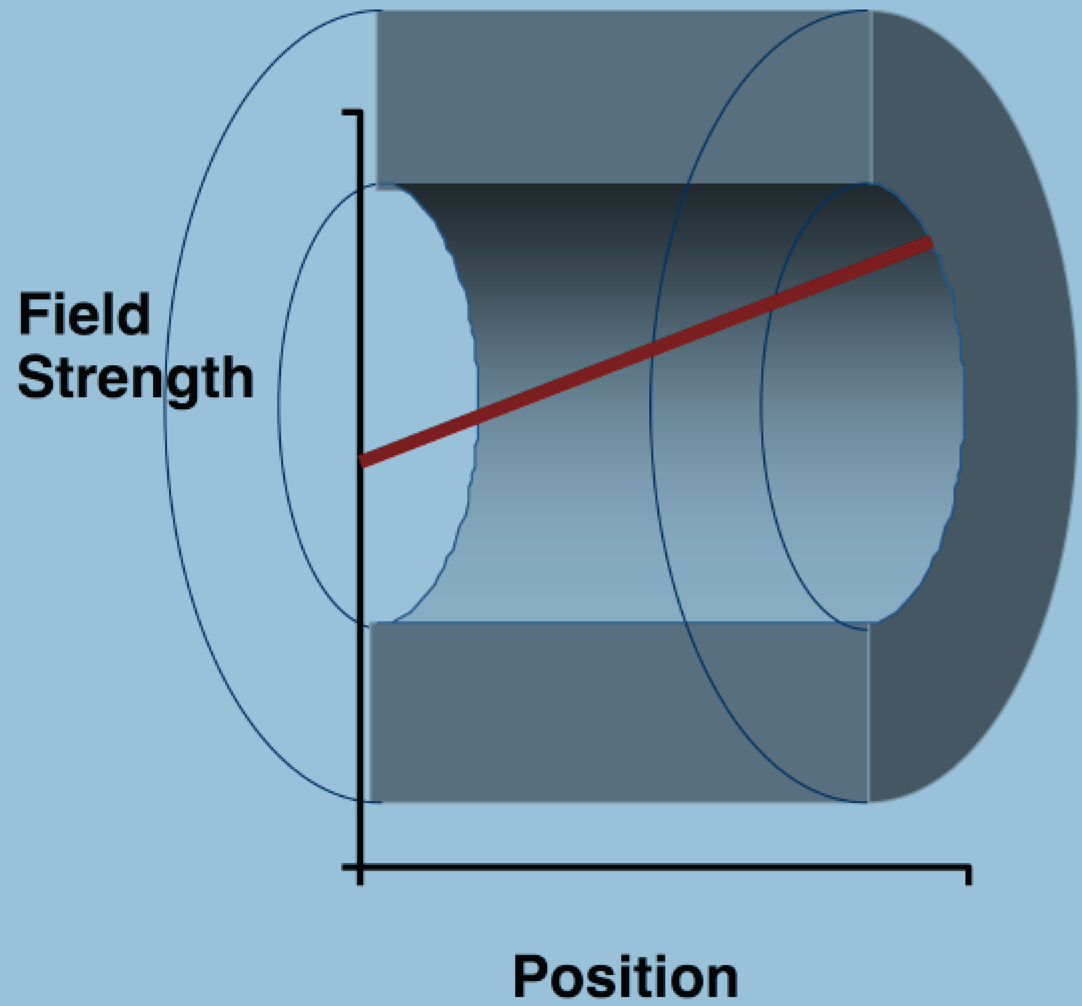


Imaging Through MR: Spin Echo

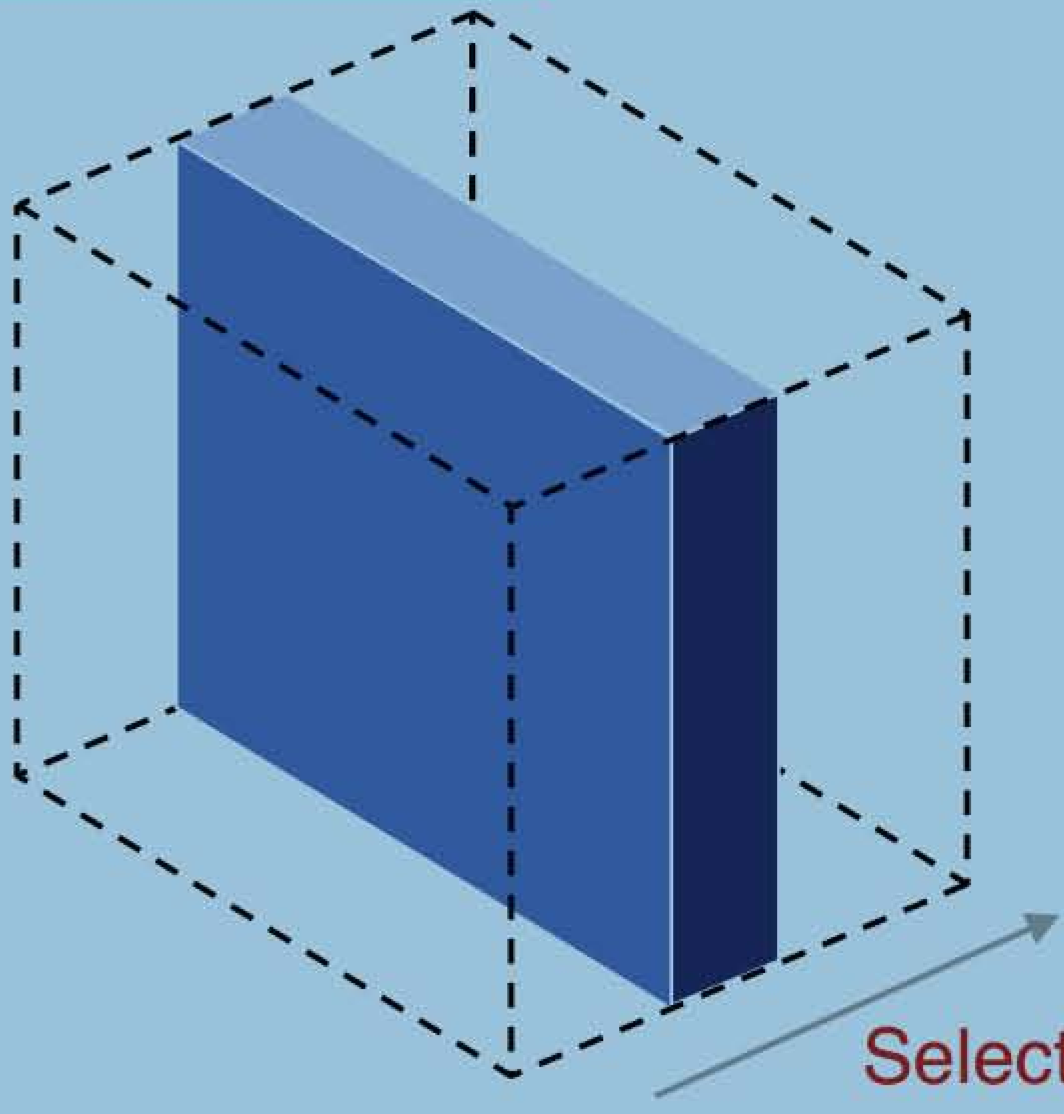




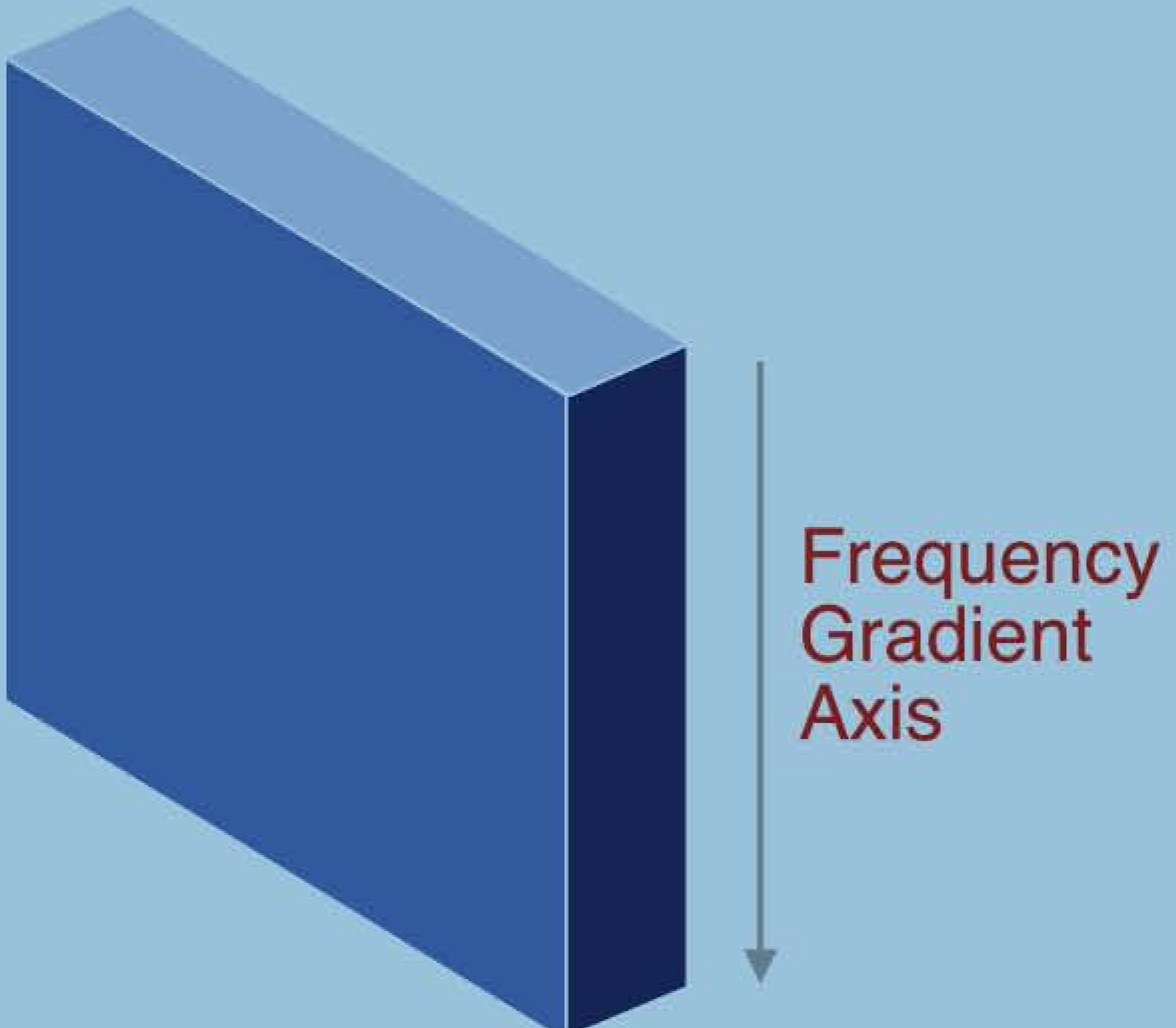
Magnetic Field Gradients



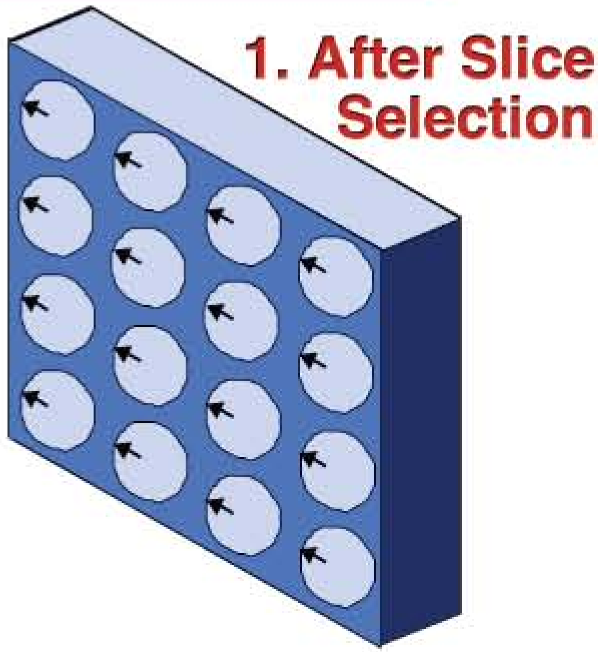
Slice Selection



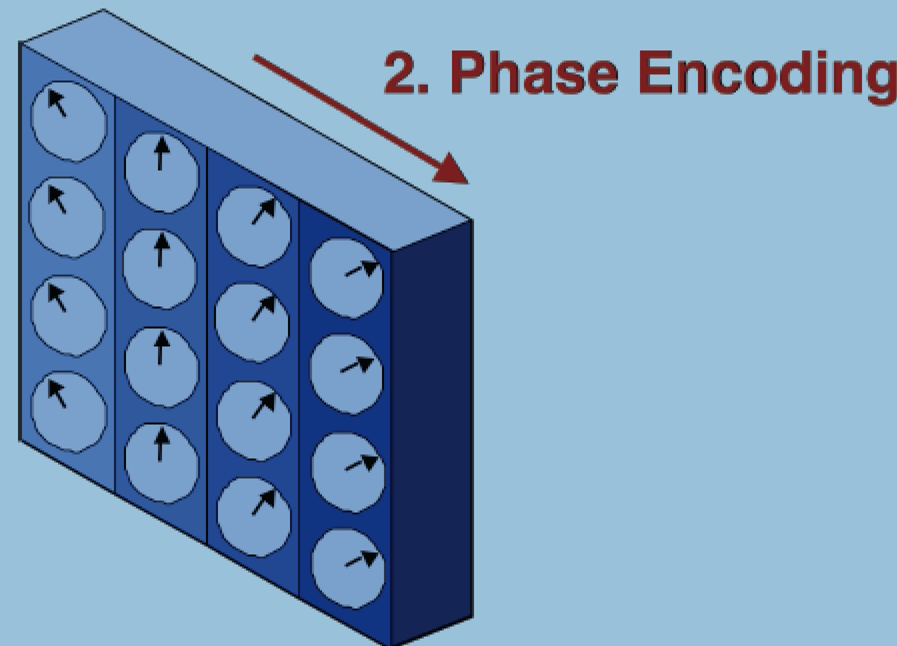
Frequency Encoding



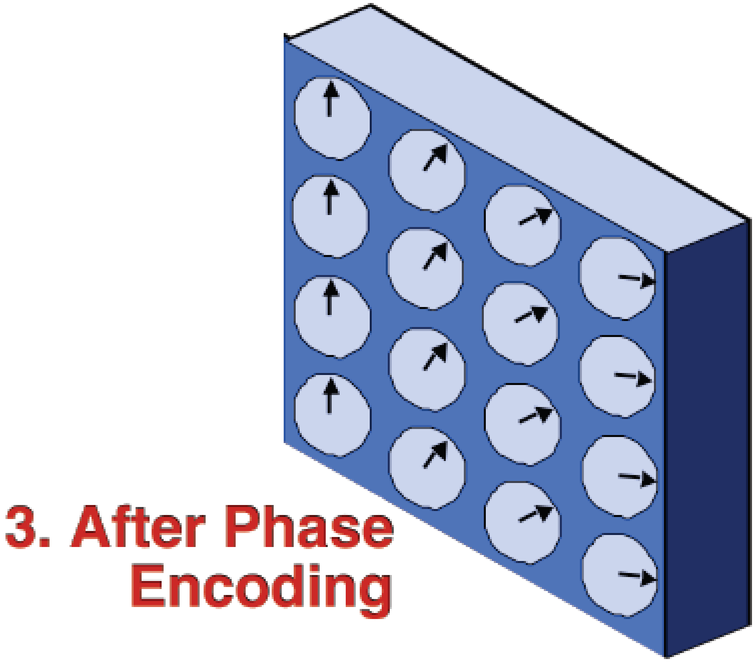
Spatial Encoding



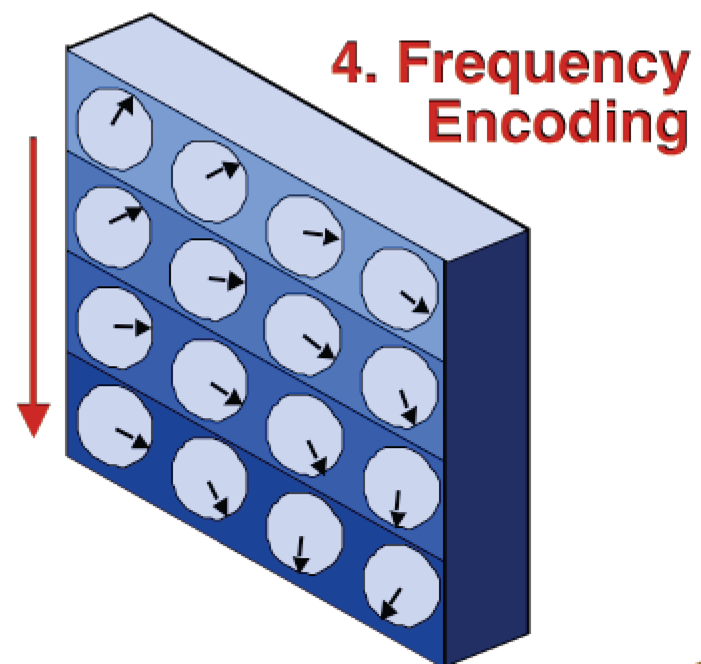
Spatial Encoding



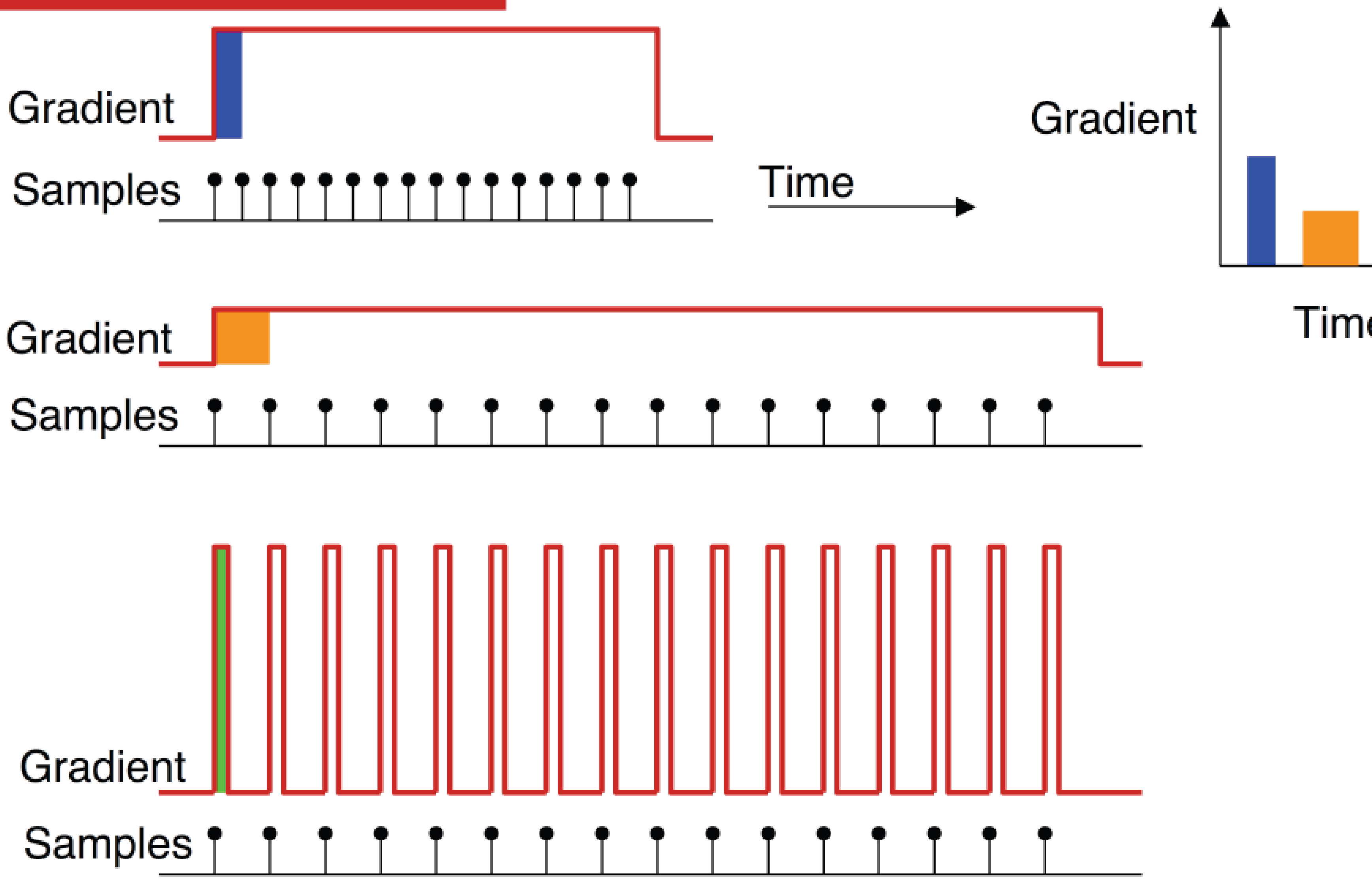
Spatial Encoding



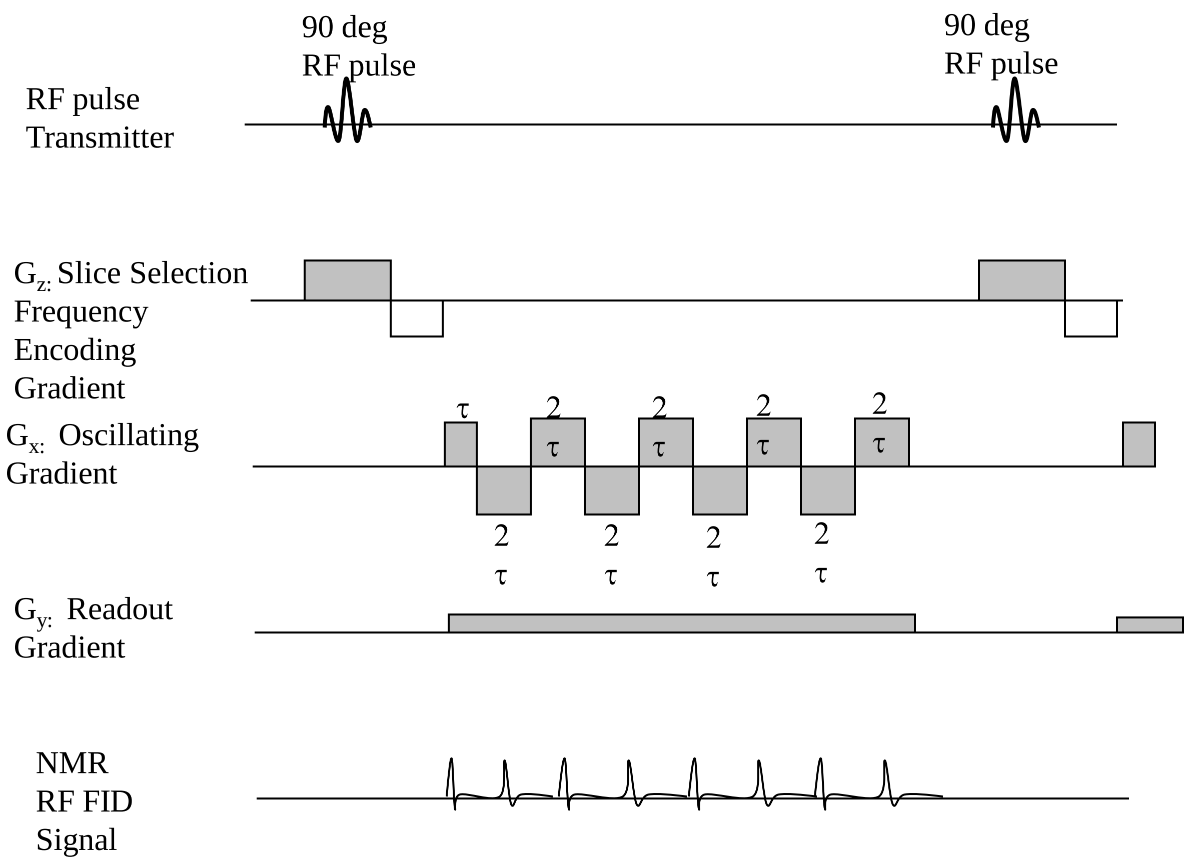
Spatial Encoding



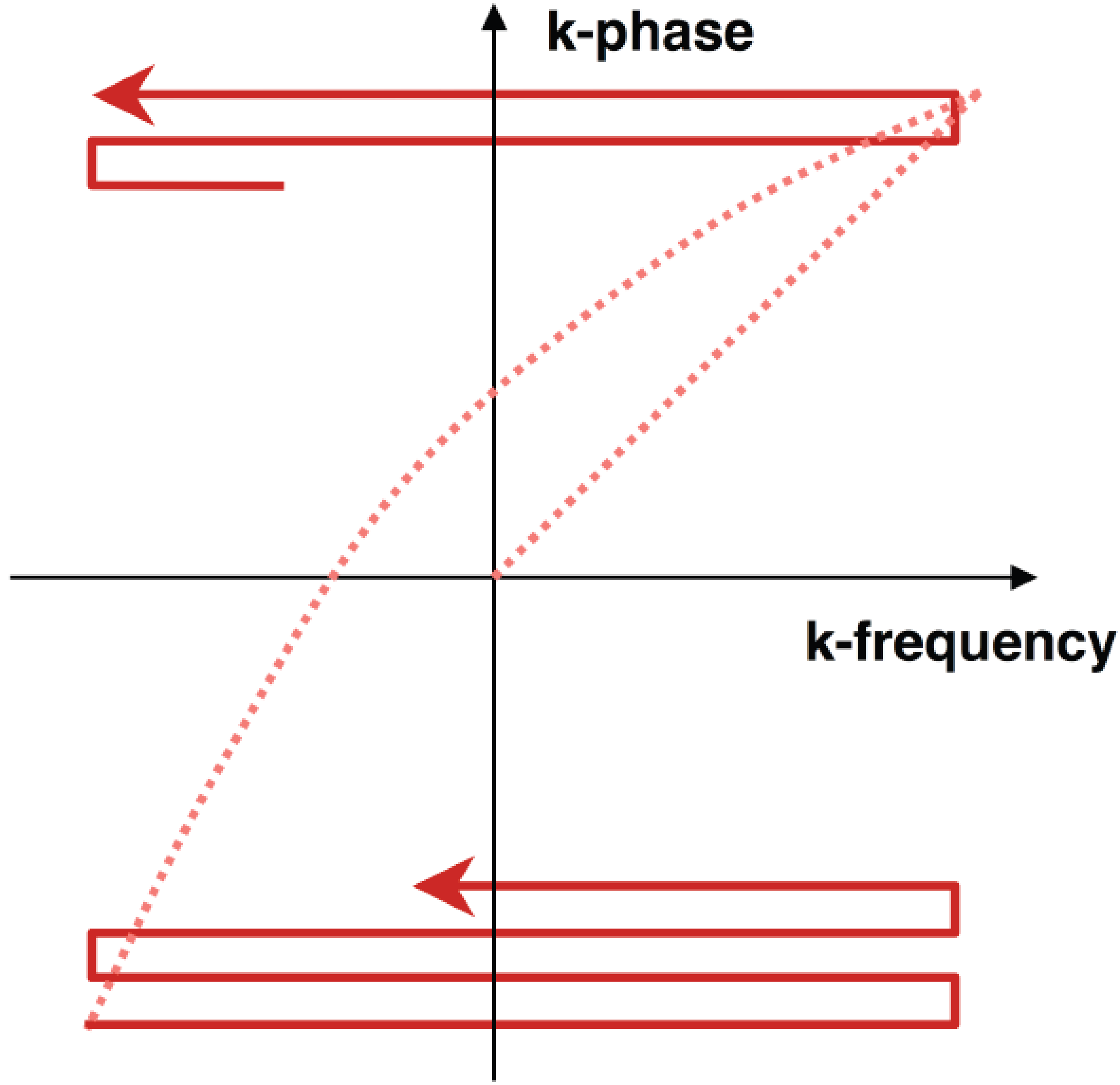
Equivalent Strategies in k-space*



MR Imaging: Single Shot EPI



Echo-Planar k-space Trajectory



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…

T1 and T2 Contrast

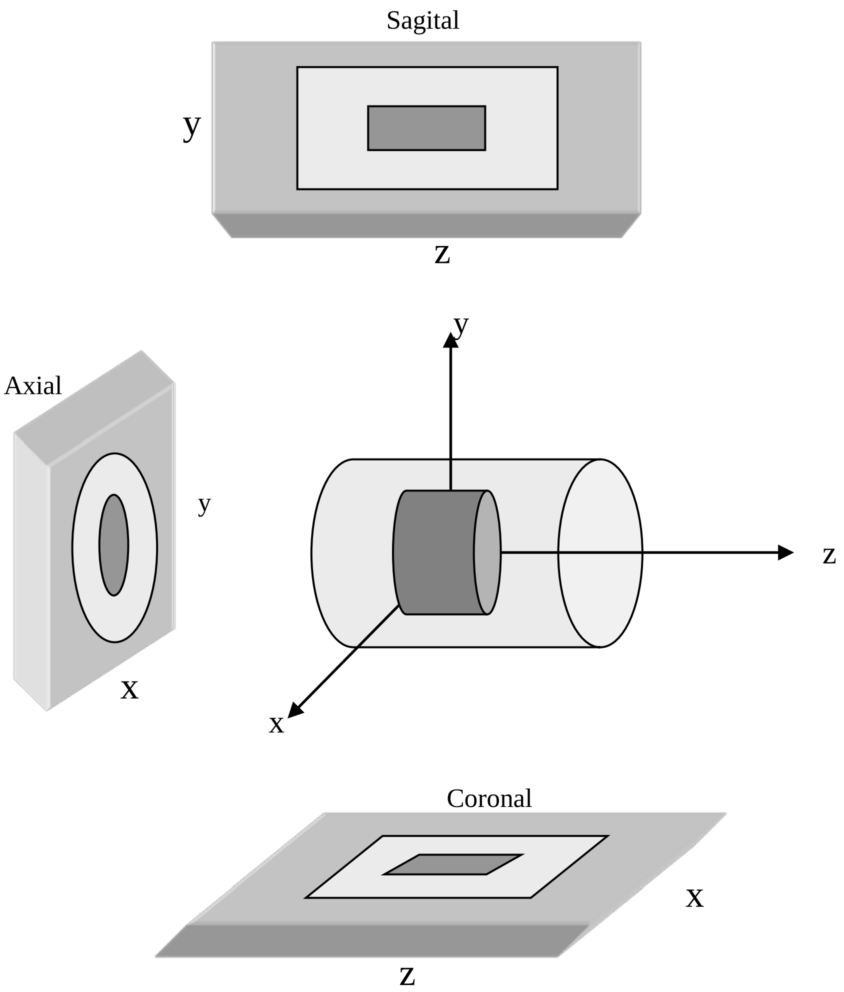
Tissue	T1 msec	T2 msec	SD %
Fat	150	150	10.9
Liver	250	44	10.0
White Matter	300	133	11.0
Gray Matter	475	118	10.5
Blood	525	261	10.0
CSF	2000	250	10.8

Typical NMR Tissue Values at 0.15 T

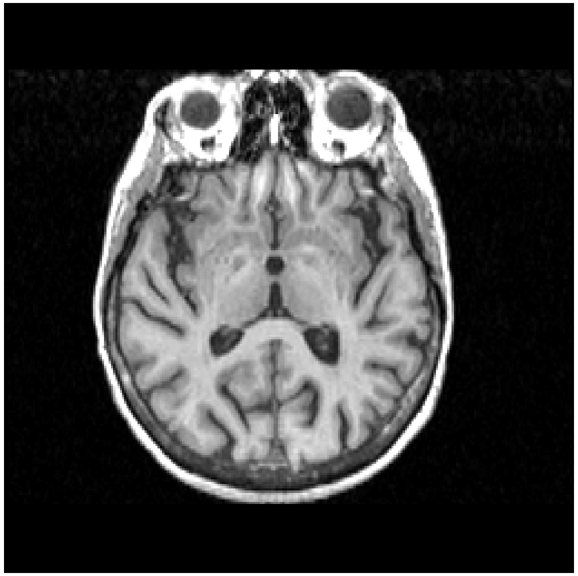
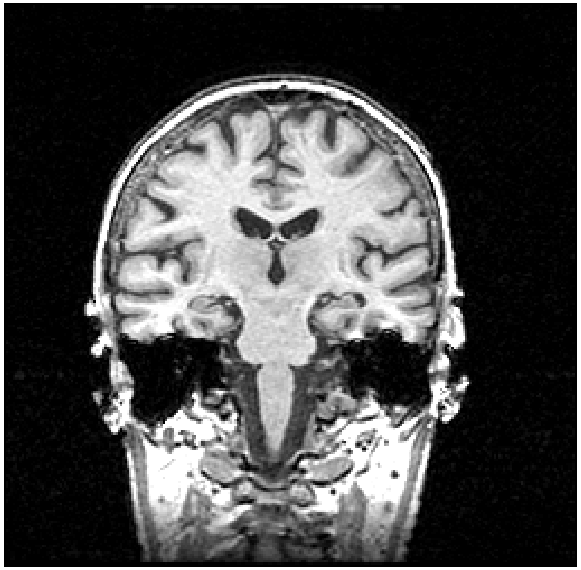
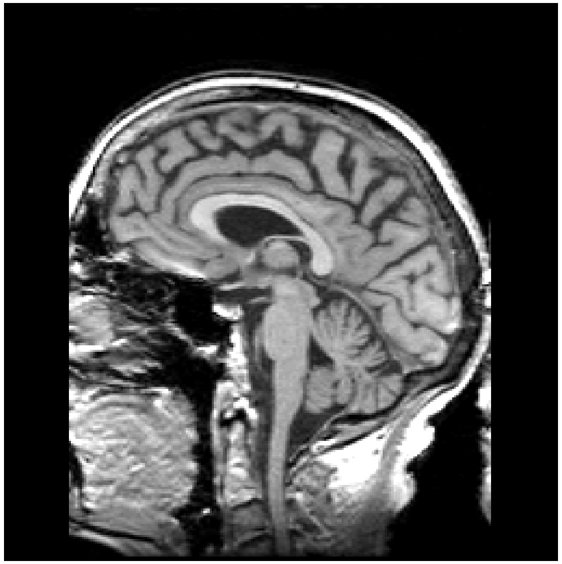
Contrast, TR and TE

TR \ TE	Short	Long
Long	Proton Density	T2-Weighted
Short	T1-Weighted	

3-D Imaging



3-D MR Imaging

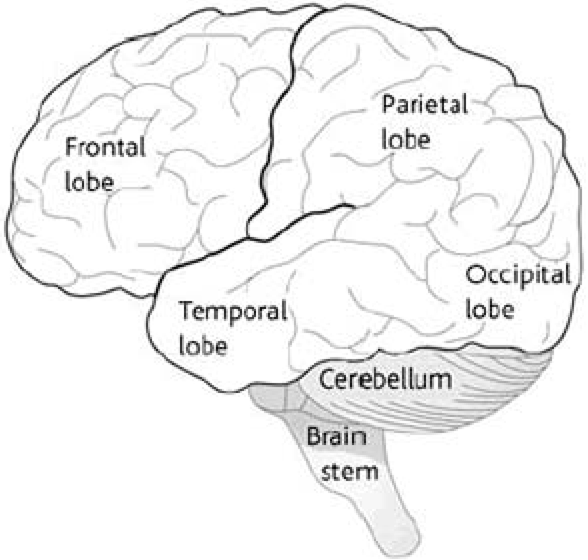
MRI

- **There are several imaging modalities within MRI.**
- **T1 and T2 weighted images**
- **Spin Echo and multiple echo sequence images**
- **MR Spectroscopy**
- **Blood flow imaging**
- **Perfusion imaging**
- **Function imaging**

MRI Advantage

- **The most important advantage of the MRI is its ability to provide unprecedented contrasts between various organs and tissues and the three-dimensional nature of imaging methods.**
- **Selective 3-D imaging is provided by appropriate selection of gradient fields and phase encoding methods.**
- **A variety of contrast images can be created by different combinations of weighting of T1, T2 and echo images**
- **MR spectroscopy provides a great potential for meaningful tissue characterization.**
- **Functional MRI holds great promise for the future.**

Advanced MRI Methods



Atam P Dhawan

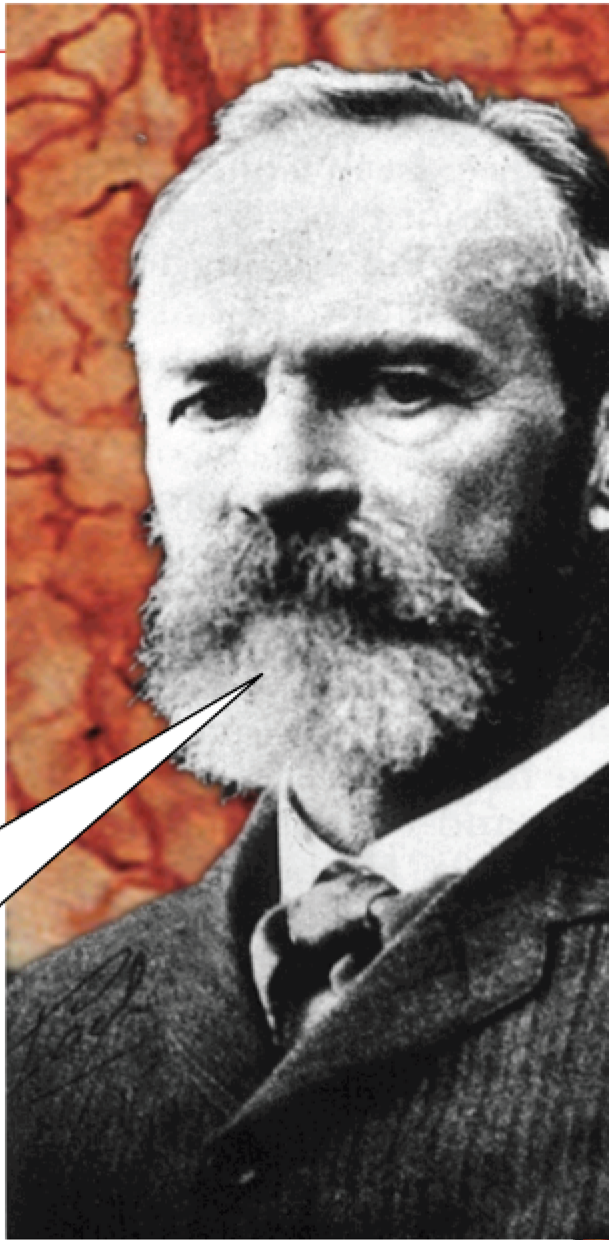
Functional MRI (fMRI)

- fMRI aims to measure the hemodynamic response related to neural activity.
  - Can measures changes in blood oxygenation levls in neural tissue (in brain or nervous systems).
  - Neural cells when active or create action potnetial consume oxygen which is taken from oxygenated hemoglobin.
  - Due to neural activity, oxygen consumption causes local changes in the relative concentration of oxyhemoglobin and deoxyhemoglobin and changes in local cerebral blood volume with an increase in blood flow.
  - Blood flow is also highly correlated with metabolic rate.

William James (1890)

"We must suppose a very delicate adjustment whereby the circulation follows the needs of the cerebral activity.

Blood very likely may rush to each region of the cortex according as it is most active, but of this we know ***nothing****."*



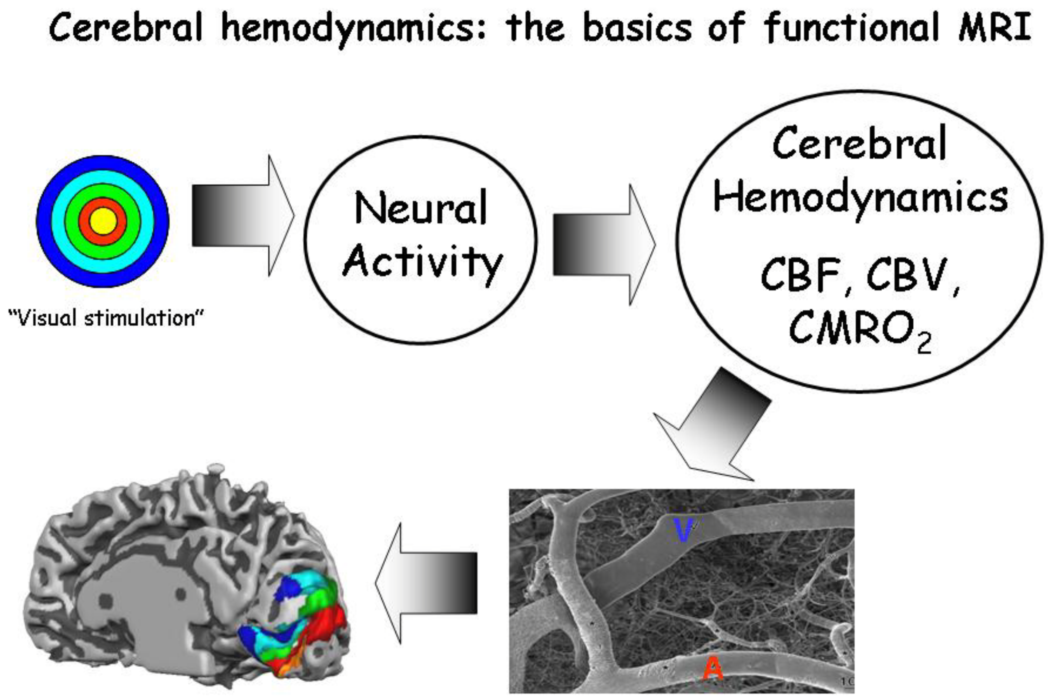
Brain "Activation" Leads to:

CBF	Increased	$+\Delta R1$
CBV	Increased	$+\Delta R2$ (C+)
O_2 Utilization	Increased slightly?	
Venous $[O_2]$	Increased	$-\Delta R2^*$
Glucose Utilization	Increased	? Lactate

$R1=1/T1$

$R2=1/T2$

Hemodynamic basis of functional MRI

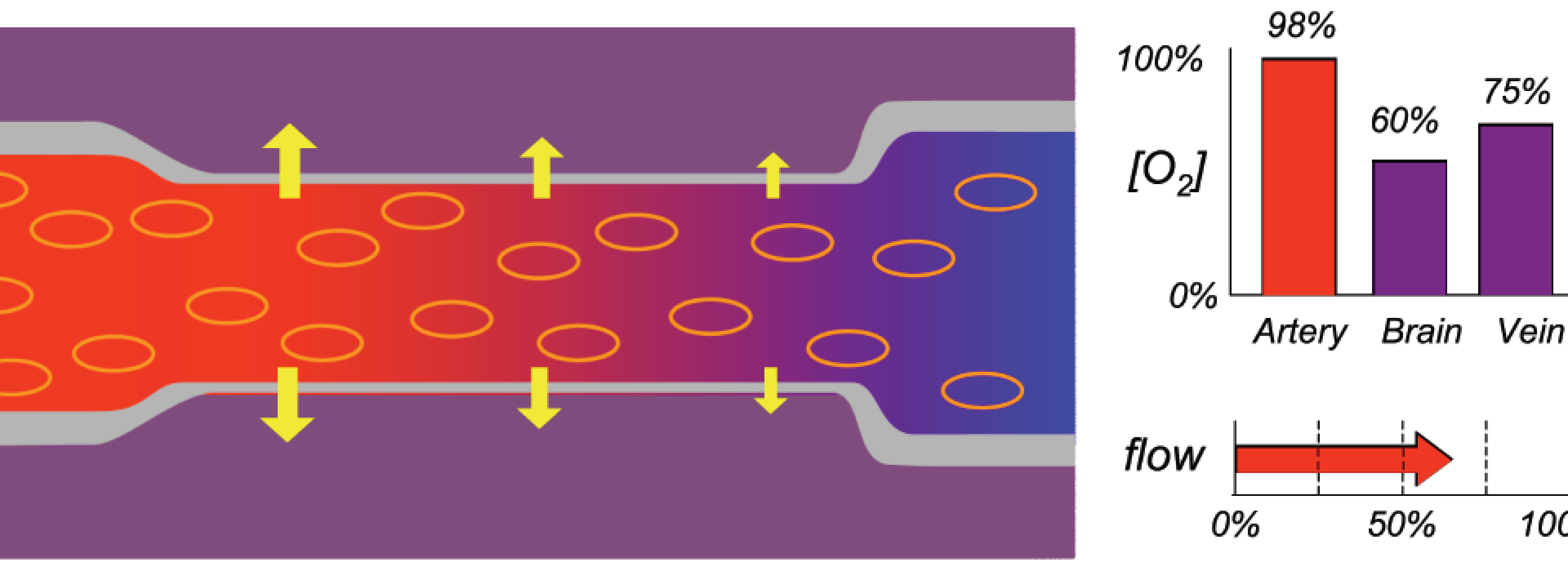


fMRI is an indirect measure of the neuronal activity elicited by an external stimulus ("visual stimulation") mediated through hemodynamic processes occurring in the dense network of veins ("V"), arteries ("A")

Blood Oxygen Level Dependent (BOLD) Imaging

- Deoxyhemoglobin as intravascular paramagnetic contrast agent:
  - Hemoglobin is diamagnetic when oxygenated but paramagnetic when deoxygenated providing different FID signals.
- Blood-Oxygen-Level Dependent (BOLD) contrast MR pulse sequence can detect level of oxygenation through deoxyhemoglobin.
- A *reduction* of the relative deoxyhemoglobin concentration due to an increase of blood flow (and hence increased supply of fresh oxyhemoglobin) during any neural or metabolic activity can be measured as an *increase* in T2 or T2 weighted MR signals.

Why Does Venous O2 Increase?

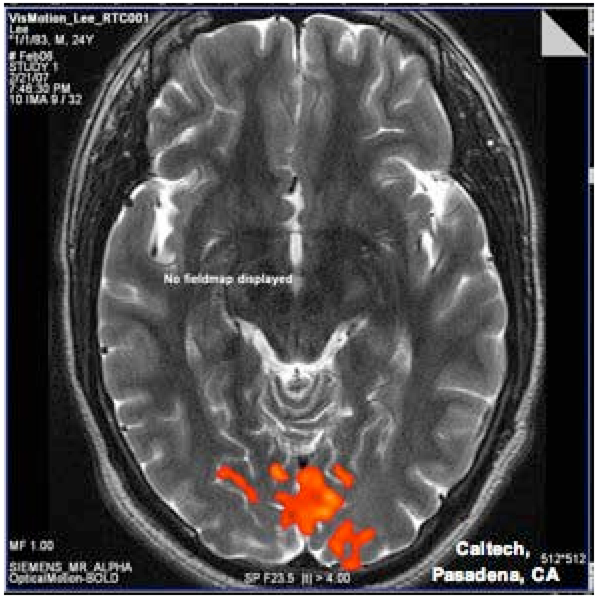


Under normal conditions oxygen diffuses down its concentration gradient from the

What BOLD Measures?

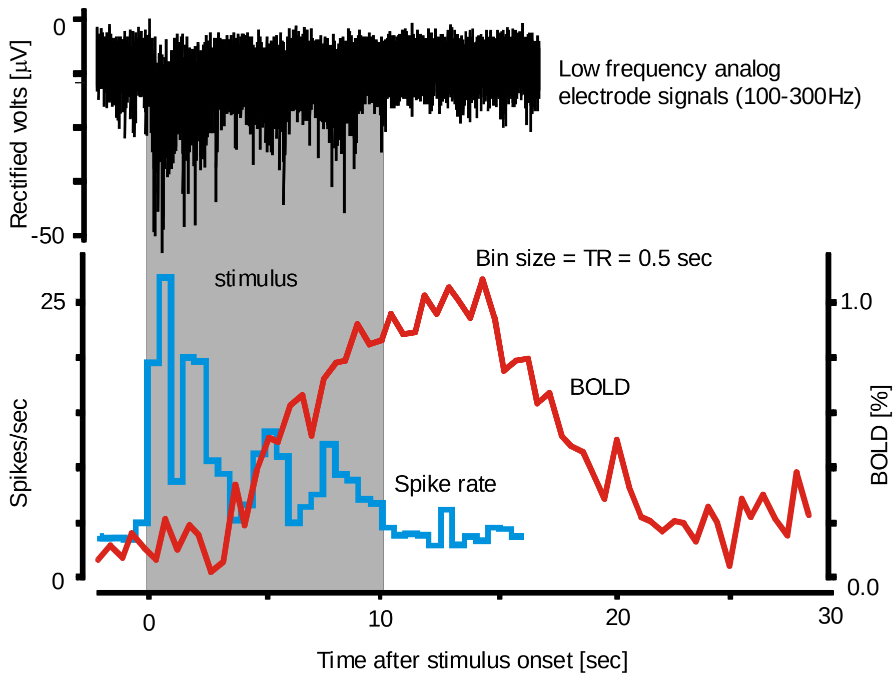
- BOLD contrast reflects a complex convolution of changes, following a neural activity, involving:
  - cerebral metabolic rate of oxygen (CMRO2)
  - cerebral blood flow (CBF), and
  - cerebral blood volume (CBV)

BOLD Contrast



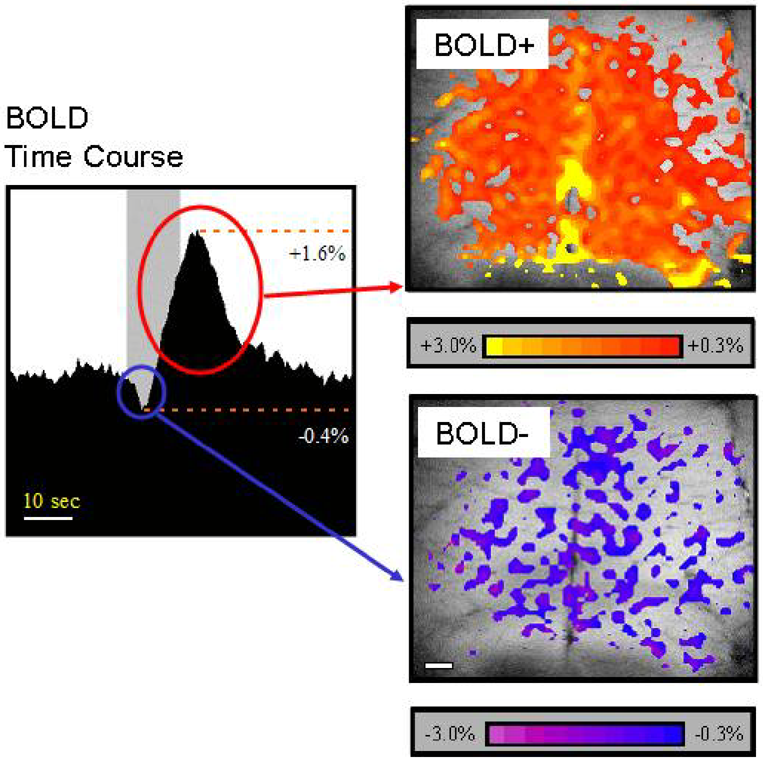
- Visual Cortex Activity with BOLD signal

BOLD Measurements



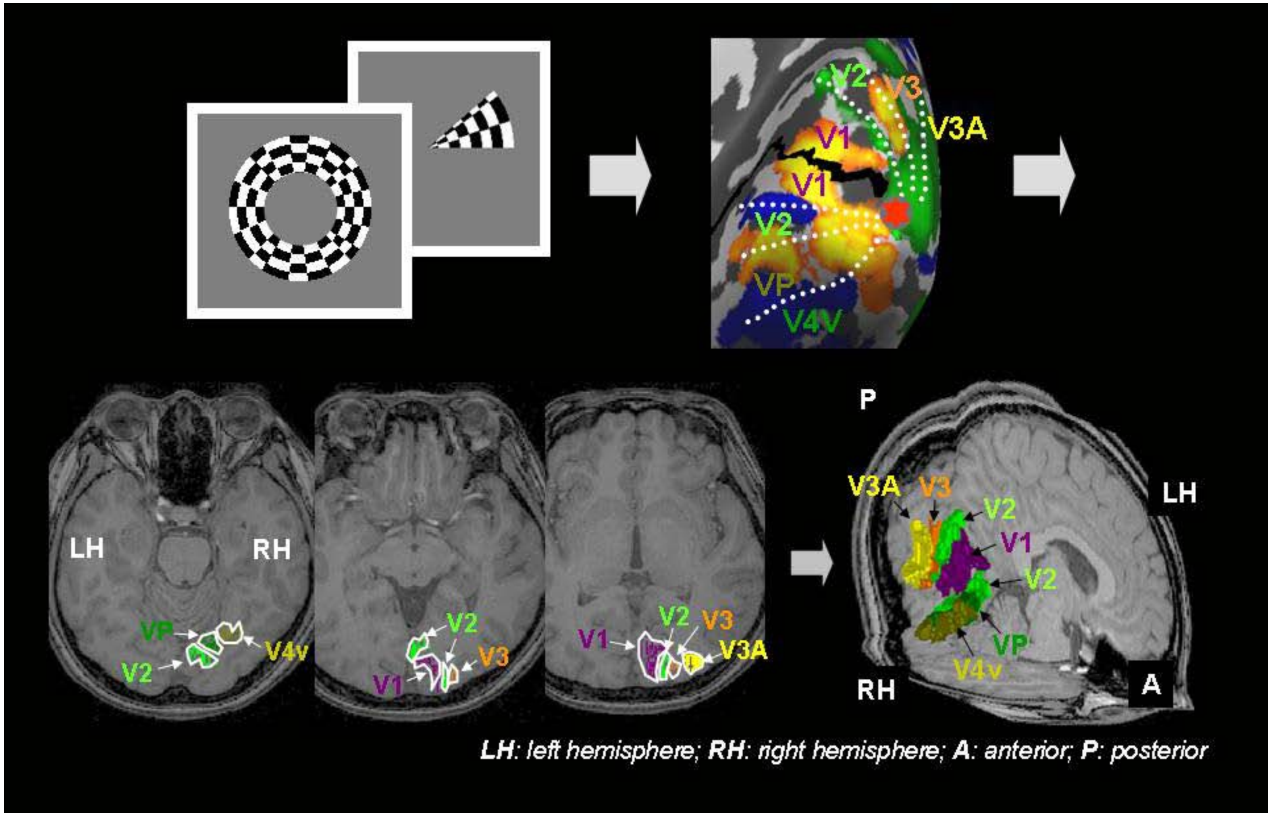
Time course of BOLD and single unit recordings from the same cortical location. Identical visual stimuli were used for fMRI and subsequent single unit recording sessions. Blue trace: peristimulus histogram of the spike activity. Red trace: BOLD percent changes during visual stimulation. Gray box: stimulus duration. The black trace above indicates the original low-frequency analog signals (100-300Hz) underlying the depicted spike counts.

BOLD Spatial Specificity



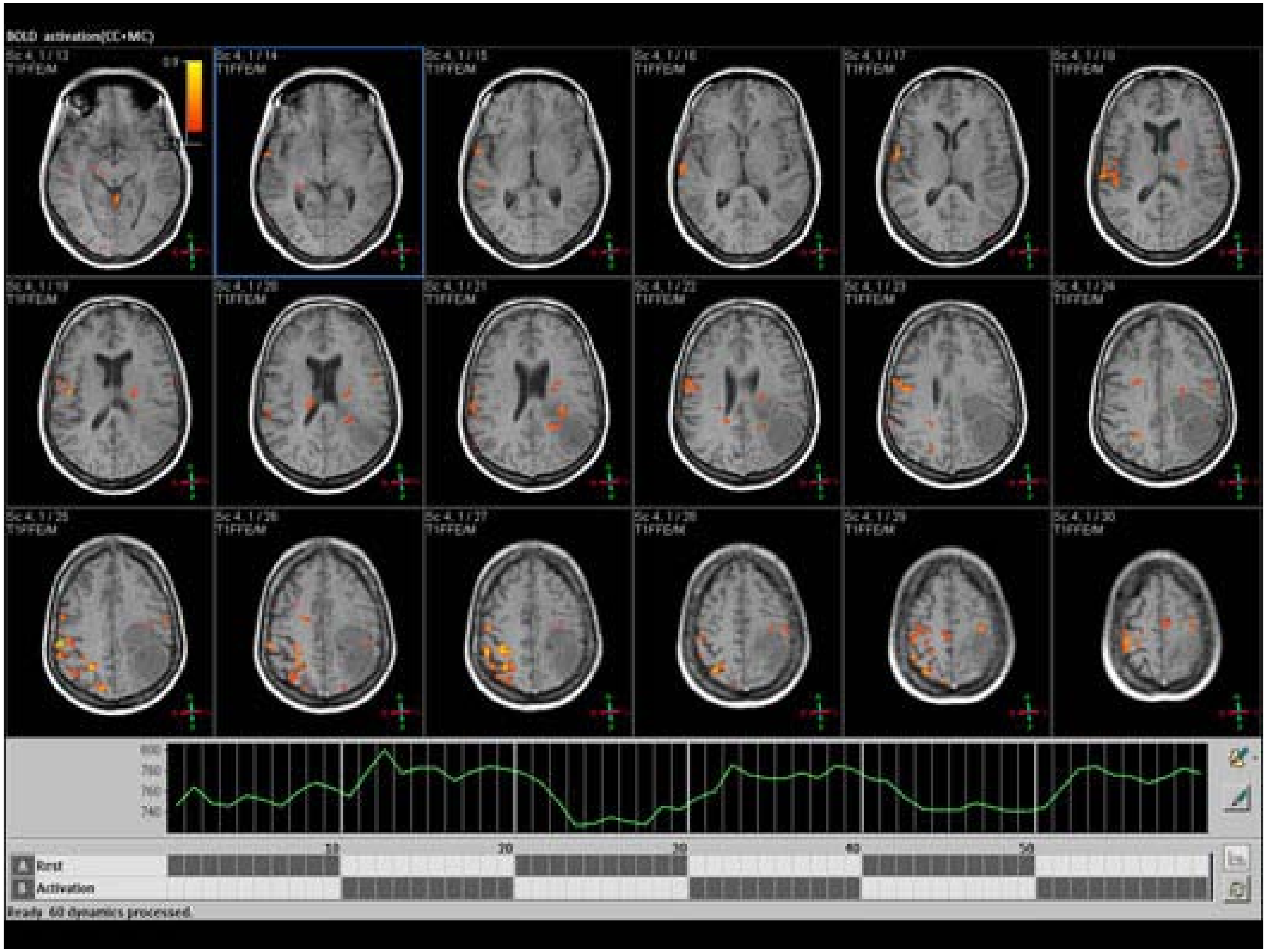
Improvement of BOLD spatial specificity by using non-conventional functional MRI signals. Time course on the left side shows biphasic evolution of MR signals, resulting the early deoxygenation contrast. If used, such deoxygenation signals produce high-resolution images of exceedingly high functional specificity (termed BOLD-) that contrasts with conventional BOLD fMRI signals (termed BOLD+).

Functional MRI of the human visual cortex: BOLD 3T



Mapping of the receptive field properties for iso-eccentricity using the standard stimuli. Color-coded activation areas were responding to eccentricities represented by the colored rings

Bold Contrast Images with Stimulus

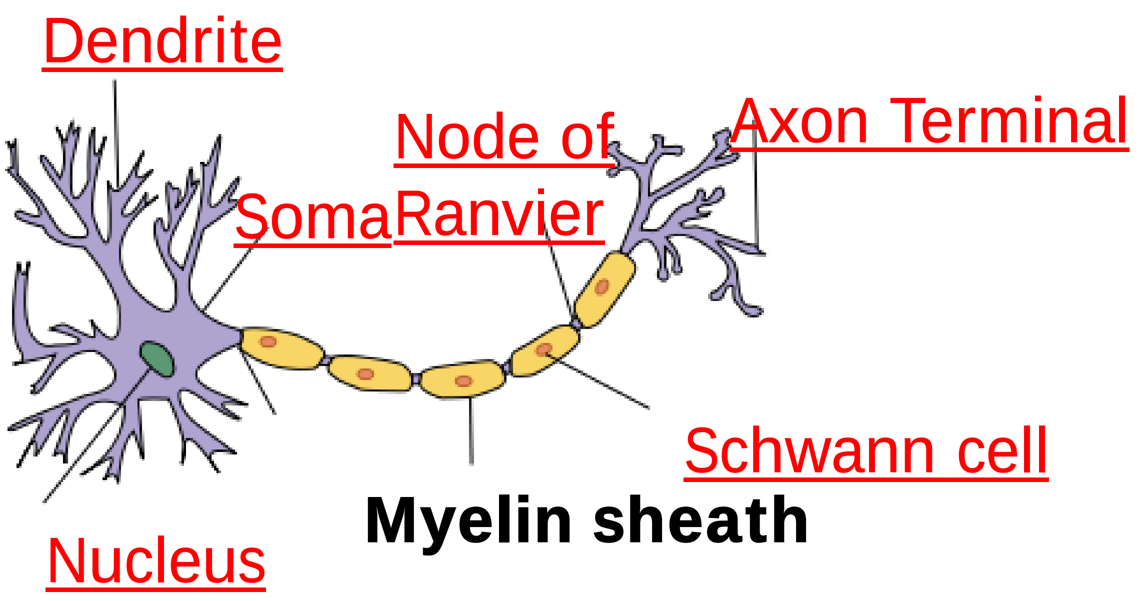


DWI/DTI

- Diffusion tensor imaging (DTI) provides information about tissue organization at the microscopic level.
- DTI probes the diffusion properties (magnitude, direction and anisotropy) of water molecules in tissues.
- The diffusion magnitude and anisotropy reflect the state of the cellular membrane permeability, myelination and axonal integrity, compartmentalization, and intrinsic and geometric hindrance to the mobility of water molecules.
- Diffusion anisotropy is related to axonal packing and axonal membranes.
- DTI allows us to visualize the location, the orientation, and the anisotropy of the brain's white matter tracts.
- Illnesses that disrupt the normal organization or integrity of cerebral white matter (such as multiple sclerosis, strokes) have a quantitative impact on DTI measures.

Anisotropic Diffusion

- The architecture of the **axons** in parallel bundles, and their **myelin** sheaths, facilitate the **diffusion** of the water molecules preferentially along their main direction. Such preferentially oriented diffusion is called *anisotropic diffusion*.



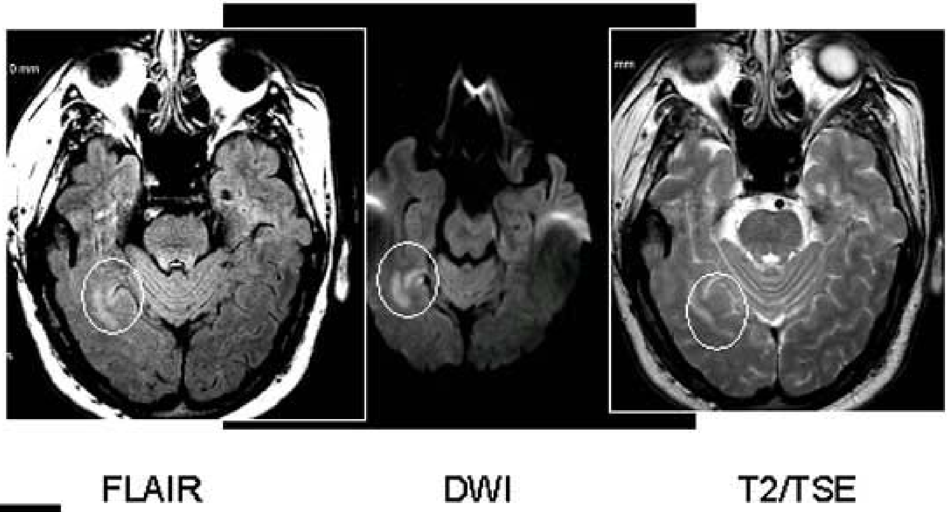
DWI

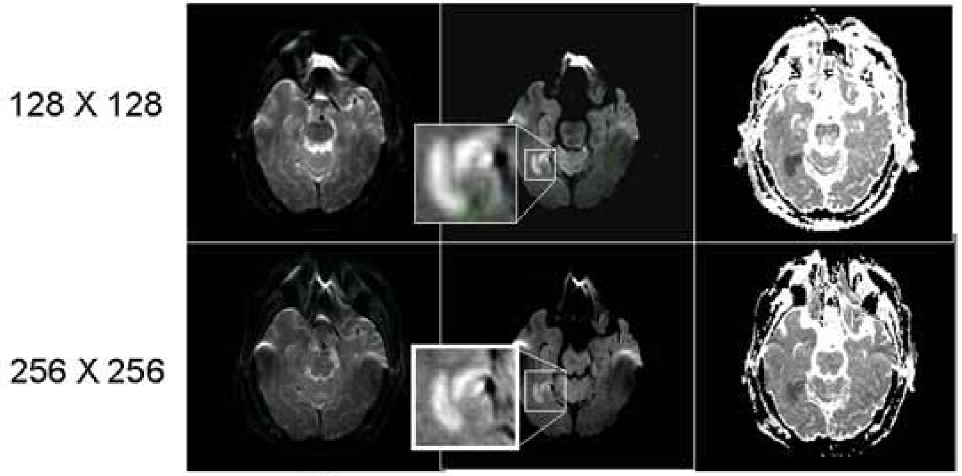
- **Diffusion-weighted imaging**

  Three gradient-directions to estimate the trace of the diffusion tensor or 'average diffusivity.

  Trace-weighted images have proven to be very useful to diagnose vascular strokes in the brain, by early detection (within a couple of minutes) of the hypoxic **edema**.

DWI Imaging Meningioma





DTI

- **Diffusion tensor imaging** (DTI) scans comprise at least six gradient directions, sufficient to compute the diffusion **tensor**.
- The diffusion model assumes homogeneity and linearity of the diffusion within each image-voxel.
- From the diffusion tensor diffusion anisotropy measures, such as the Fractional Anisotropy (FA), can be computed.
- The principal direction of the diffusion tensor can be used to infer the white-matter connectivity of the brain (**tractography**).

DTI Applications

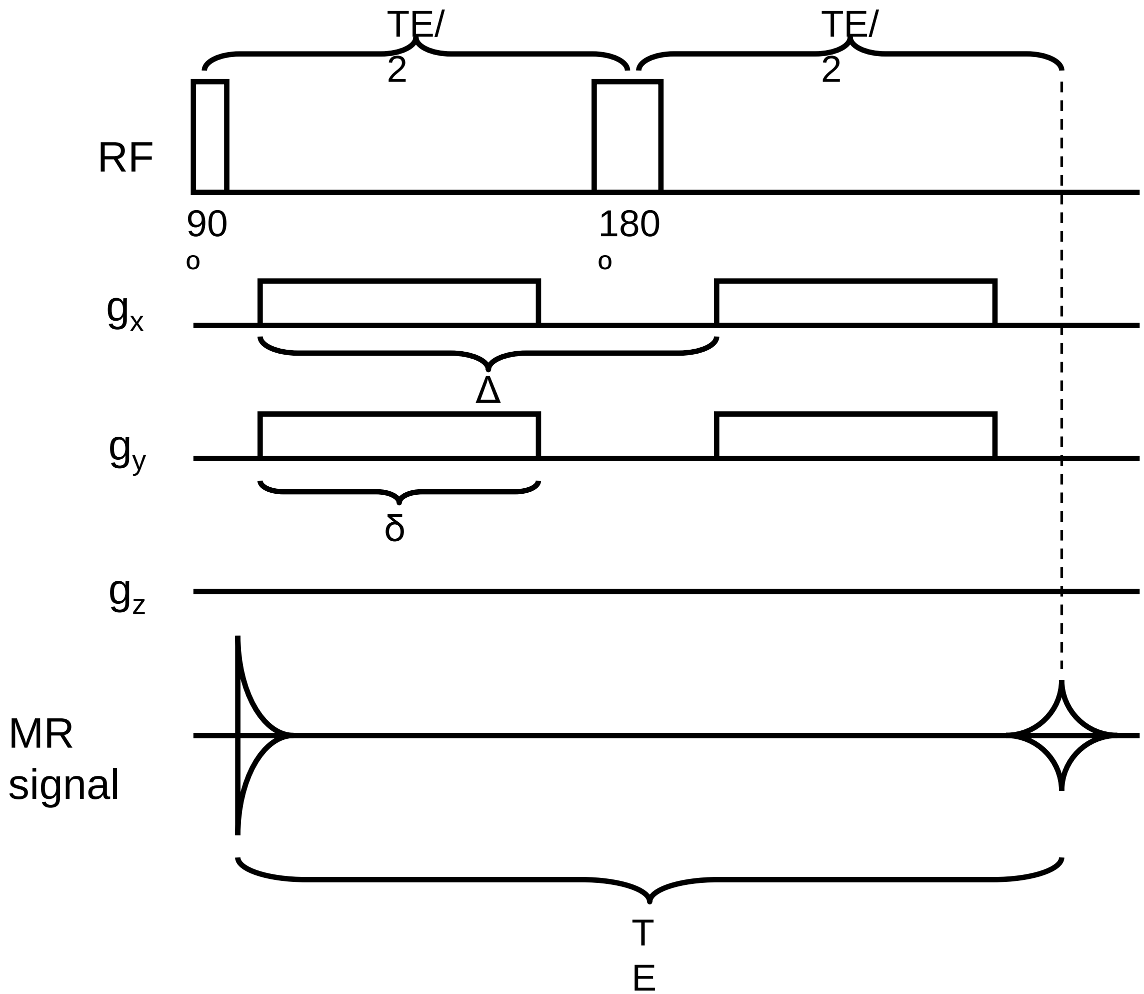
- DTI is useful to study diseases of the white matter and connectivity of brain pathways.
  - Attention deficit hyperactivity disorder (ADHD)
    - Observed abnormalities of the fiber pathways in the frontal cortex, basal ganglia, brain stem and cerebellum.
  - Schizophrenia
    - Observed abnormalities in two functionally and anatomically different neural pathways – the uncinate fasciculus (UF) and the cingulate bundle (CB).
  - Vascular Strokes
    - DTI is useful to diagnose vascular strokes in the brain, study diseases of the white matter and to see connectivity of the brain.

Diffusion

- Water molecules that start at the same location spreads out over time. Each molecule experience a series of random displacements so that after a time T the spread of position along a spatial axis *x* has a variance of $\sigma_x^2 = 2DT$ where *D* is the diffusion coefficient.

$$\sigma_x^2 = 2DT$$

DTI Pulse Sequence: g=(1,1,0)



$$\frac{S}{S_0} = e^{-\gamma^2 G^2 \delta^2 (\Delta - \delta/3) D}$$

G and δ are gradient strength and duration, and Δ is the separation between a pair of gradient pulses

DTI Measurement

$$\frac{S}{S_0} = e^{-\gamma^2 G^2 \partial^2 (\Delta - \partial/3) D}$$

$$b = \gamma^2 G^2 \partial^2 (\Delta - \partial / 3) D$$

$$S = S_0 \exp(-bD)$$

$$D = \frac{1}{b} \ln \frac{S_0}{S}$$

D is scalar in DWI but is tensor in DTI described by directio

Directional Gradient: Example

- If the diffusion-sensitizing gradient pulses are applied along the *x*-axis, **u** = (1, 0,0), or if the measurement axis is at an angle θ to the x-axis and in the *x*-*y* plane, u = (cos θ, sin θ, 0), then the measured value of *D* along any axis u is given by:

$$D = \begin{pmatrix} u_x & u_y & u_z \end{pmatrix} \begin{bmatrix} D_{xx} & D_{xy} & D_{xz} \\ D_{xy} & D_{yy} & D_{yz} \\ D_{xz} & D_{yz} & D_{zz} \end{bmatrix} \begin{pmatrix} u_x \\ u_y \\ u_z \end{pmatrix}$$

$$D = {u_x}^2 D_{xx} + {u_y}^2 D_{yy} + {u_z}^2 D_{zz} + 2u_x u_y D_{xy} + 2u_y u_z D_{yz} + 2u_z u_x D_{zx}$$

Diffusion Signal

$$D = \frac{1}{b} \ln \frac{S_0}{S}$$

$$D = u_x{}^2 D_{xx} + u_y{}^2 D_{yy} + u_z{}^2 D_{zz} + 2u_x u_y D_{xy} + 2u_y u_z D_{yz} + 2u_z u_x D_{zx}$$

$$\therefore \frac{1}{b} \ln \frac{S_0}{S} = u_x{}^2 D_{xx} + u_y{}^2 D_{yy} + u_z{}^2 D_{zz} + 2u_x u_y D_{xy} + 2u_y u_z D_{yz} + 2u_z u_x D_{zx}$$

Example 12 Directions

$$\begin{bmatrix} \frac{1}{b}\ln\frac{S_0}{S_1} \\ \cdot \\ \cdot \\ \cdot \\ \cdot \\ \frac{1}{b}\ln\frac{S_0}{S_{12}} \end{bmatrix} = U\vec{D}$$

$$U = \begin{bmatrix} u_{x1}{}^2 & u_{y1}{}^2 & u_{z1}{}^2 & u_{x1}u_{y1} & u_{y1}u_{z1} & u_{z1}u_{x1} \\ \cdot & \cdot & \cdot & \cdot & \cdot & \cdot \\ \cdot & \cdot & \cdot & \cdot & \cdot & \cdot \\ \cdot & \cdot & \cdot & \cdot & \cdot & \cdot \\ \cdot & \cdot & \cdot & \cdot & \cdot & \cdot \\ u_{x12}{}^2 & u_{y12}{}^2 & u_{z12}{}^2 & u_{x12}u_{y12} & u_{y12}u_{z12} & u_{z12}u_{x12} \end{bmatrix}$$

$$\vec{D} = \begin{bmatrix} D_{xx} \\ D_{yy} \\ D_{zz} \\ 2D_{xy} \\ 2D_{yz} \\ 2D_{zx} \end{bmatrix}$$

Now, if we assume that the columns of U are linearly independent, then the matrix UTU is invertible and the least squares solution is

$$\vec{D}_0 = (U^T U)^{-1} U^T \begin{bmatrix} \frac{1}{b}\ln\frac{S_0}{S_1} \\ \cdot \\ \cdot \\ \cdot \\ \cdot \\ \frac{1}{b}\ln\frac{S_0}{S_{12}} \end{bmatrix}$$

Tensor Matrix and Eigenvalues

The 3x3 tensor matrix

$$\overline{\overline{D}} = \begin{bmatrix} D_{xx} & D_{xy} & D_{xz} \\ D_{xy} & D_{yy} & D_{yz} \\ D_{xz} & D_{yz} & D_{zz} \end{bmatrix}$$

is symmetric along the diagonal.
The eigenvalues and eigenvectors can be obtained by diagonalizing the matrix using the Jacobi transformation.
The resulting eigenvalues

$$\overline{\overline{\Lambda}} = \begin{bmatrix} \lambda_1 & 0 & 0 \\ 0 & \lambda_2 & 0 \\ 0 & 0 & \lambda_3 \end{bmatrix}$$

and corresponding eigenvectors $\overline{\overline{P}} = \begin{bmatrix} \overrightarrow{p_1} & \overrightarrow{p_2} & \overrightarrow{p_3} \end{bmatrix}$

can then be used to describe the diffusivity and directionality (or anisotropy) of water diffusion within a given voxel.

An important measure associated with the diffusion tensor is its trace:

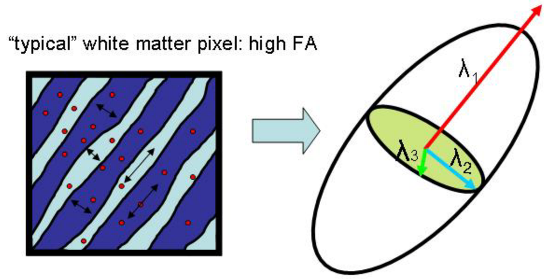
$$tr\{\underline{D}\} = D_{xx} + D_{yy} + D_{zz} = 3 \cdot \langle \lambda \rangle = \lambda_1 + \lambda_2 + \lambda_3$$

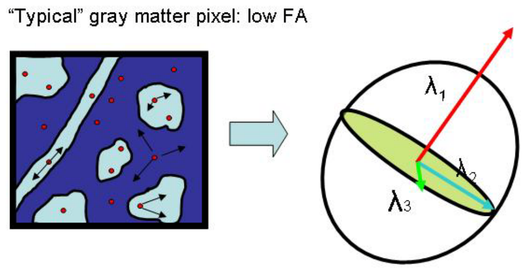
Fractional Anisotropy (FA)

The fractional anisotropy (FA) (Basser and Pierpaoli 1996):

$$FA = \frac{1}{\sqrt{2}} \sqrt{\frac{(\lambda_1 - \lambda_2)^2 + (\lambda_2 - \lambda_3)^2 + (\lambda_3 - \lambda_1)^2}{\lambda_1{}^2 + \lambda_2{}^2 + \lambda_3{}^2}}$$

Diffusion Ellipsoid



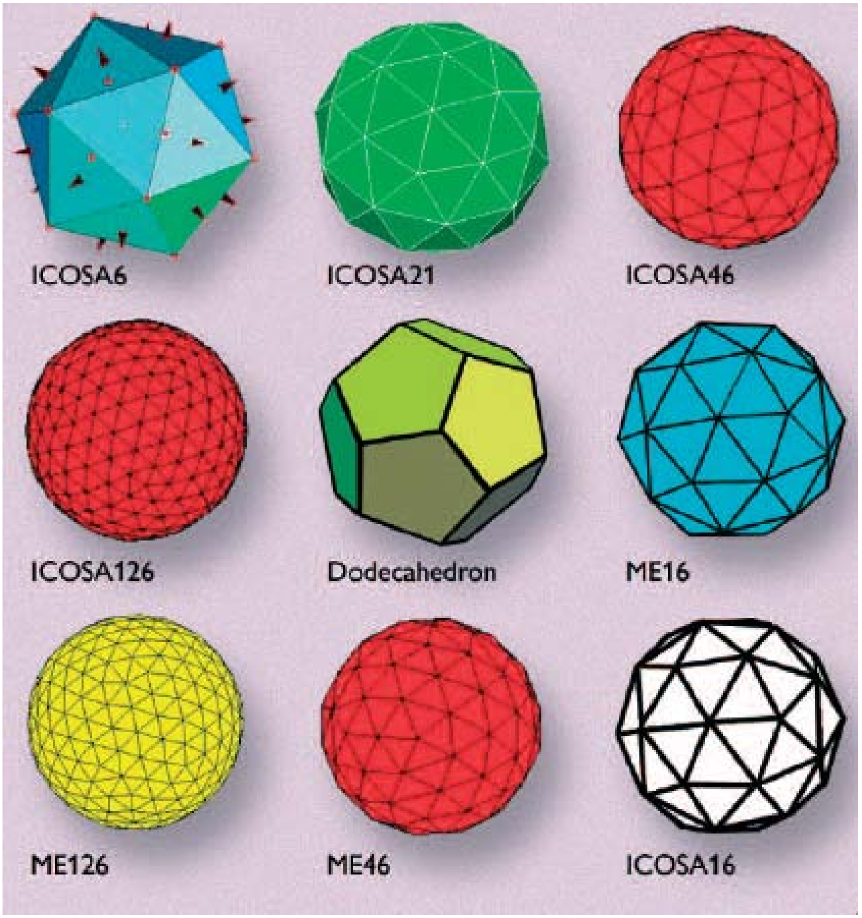


$\lambda 1 >> \lambda 2 \geq \lambda 3$ (*anisotropic* diffusion)
$\lambda 1 \approx \lambda 2 \approx \lambda 3$ (isotropic diffusion)
In anisotropic diffusion, $\lambda 1$ indicates the direction of fiber. Isotropic diffusion suggests unaligned fibers.

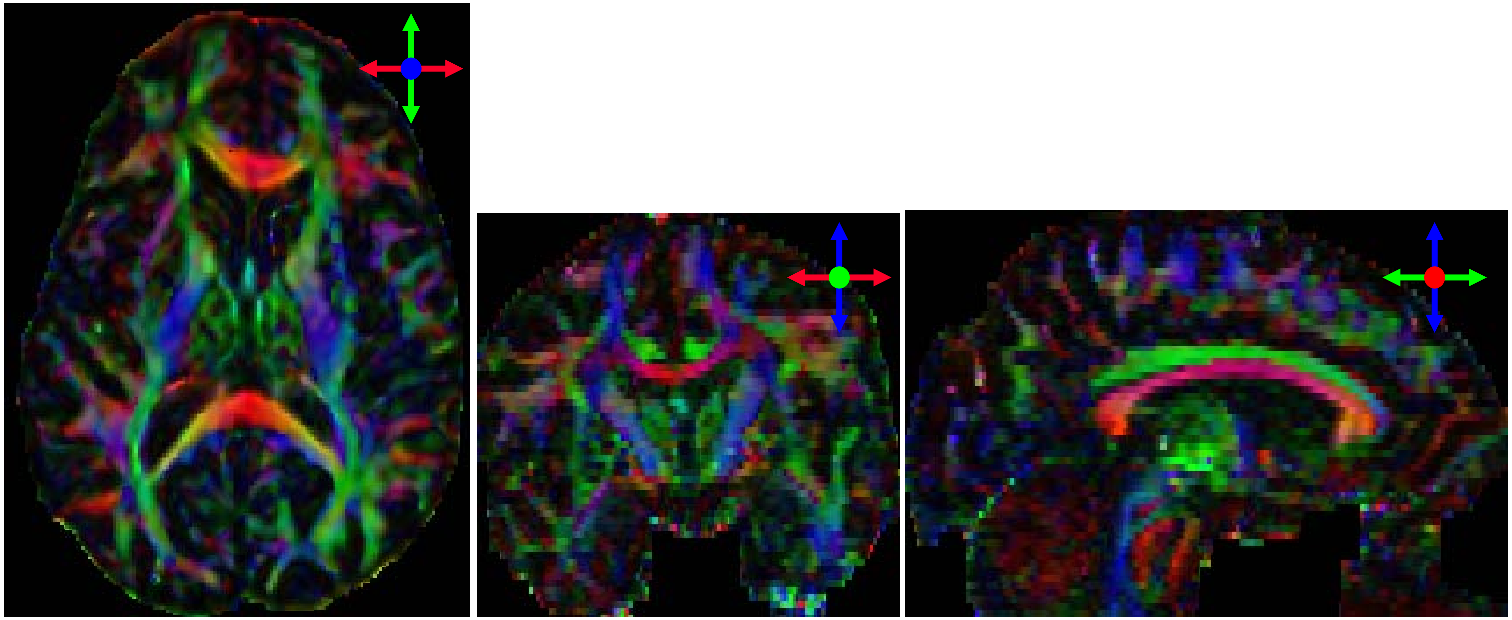
White matter voxel is mostly occupied by closely packed myelinated axons. Water molecule diffusion is restricted in the direction perpendicular to the axonal fibers leading to an anisotropic diffusion pattern.

In a gray-matter voxel, although the presence of cell membranes still poses restritcion on diffusion, the well-oriented structure of white matter fiber tract no longer exists, and thus the diffusion pattern is more isotropic.

Isotropically Distributed Tensor Encoding Sets



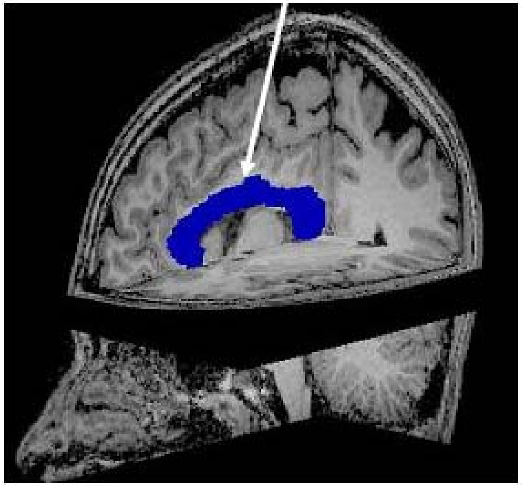
Eigenvalues Color Maps

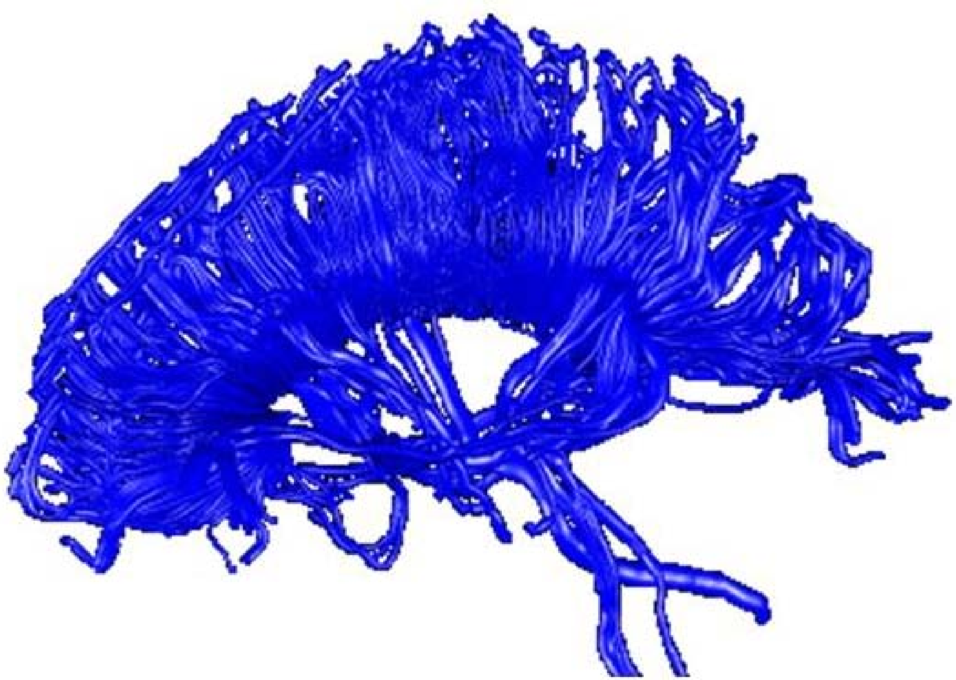


- The fibers that are oriented from left to right of the brain appear red, the fibers oriented anteriorly-posteriorly (front-back) appear green, and those oriented superiorly-inferiorly (top-bottom) appear blue.

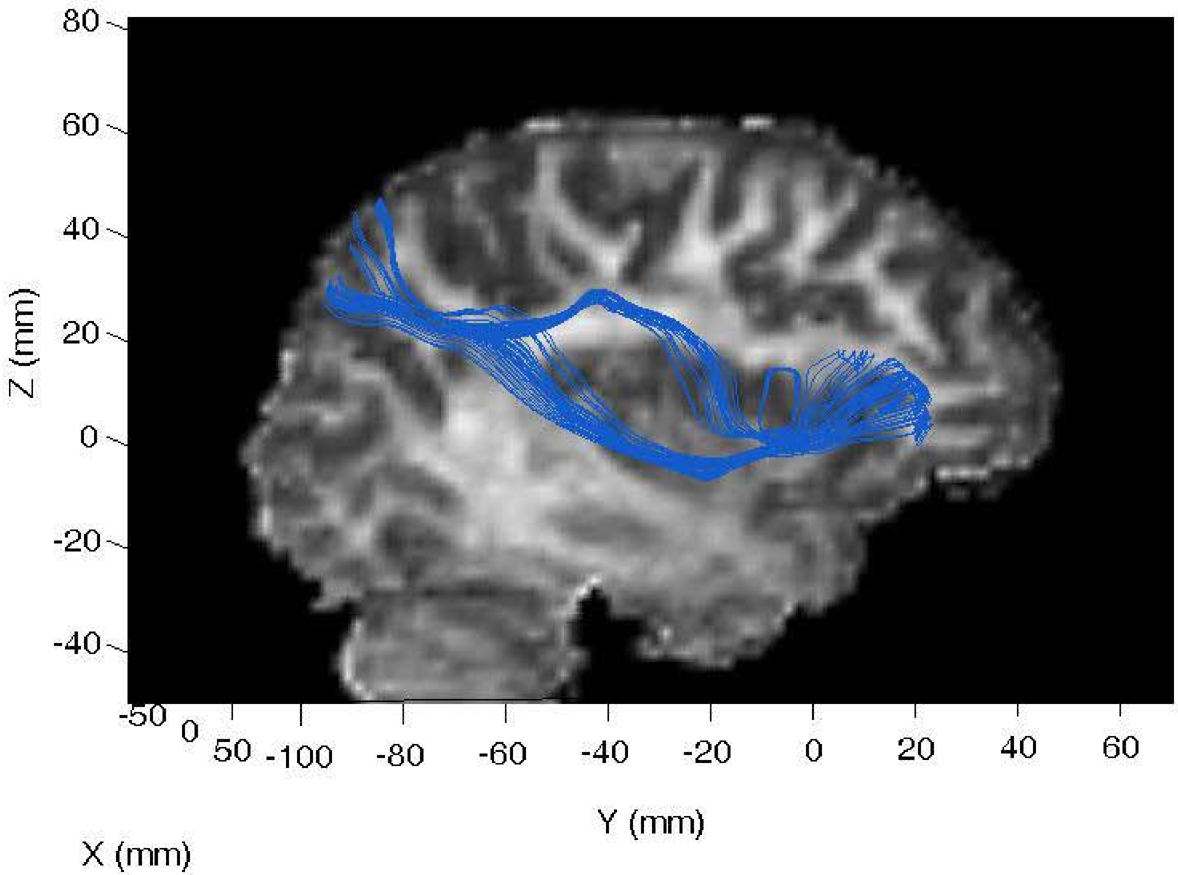
Fiber tractography of human corpus callosum.

Seeding ROI





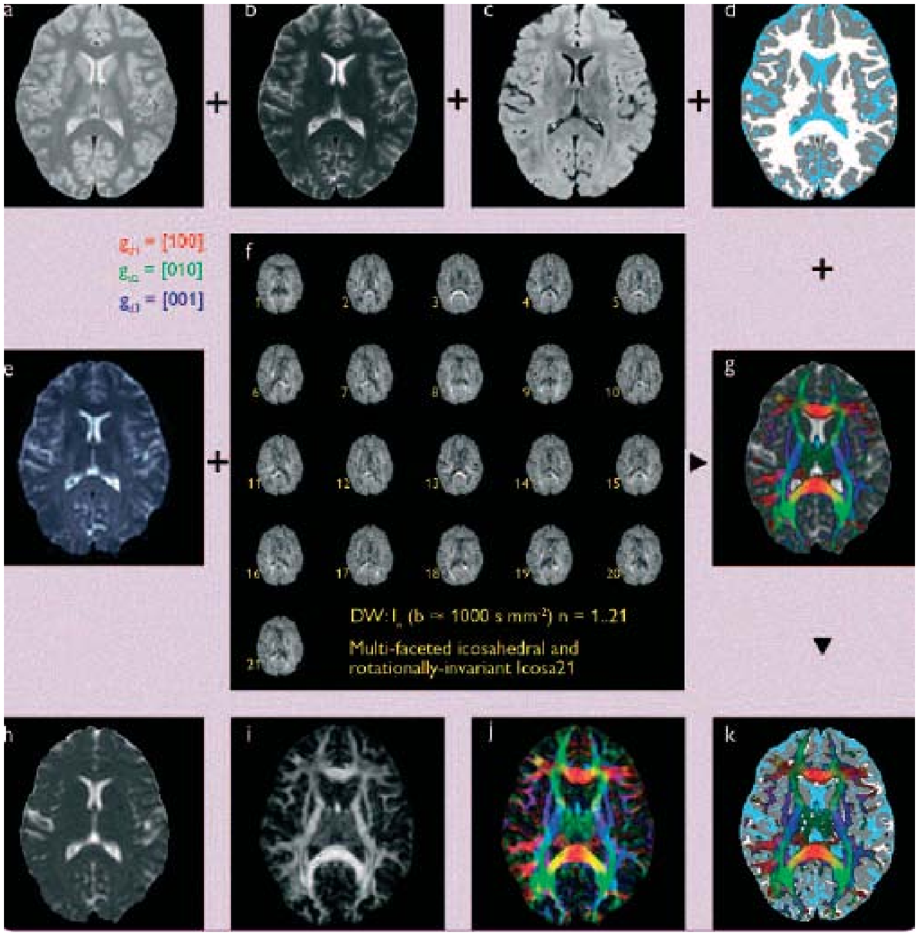
Fiber-trajectories of Corpus Callosum


80
60
40
20
0
-20
-40
Z (mm)
-50
0
50
-100
-80
-60
-40
-20
0
20
40
60
Y (mm)
X (mm)

DTI with FLAIR

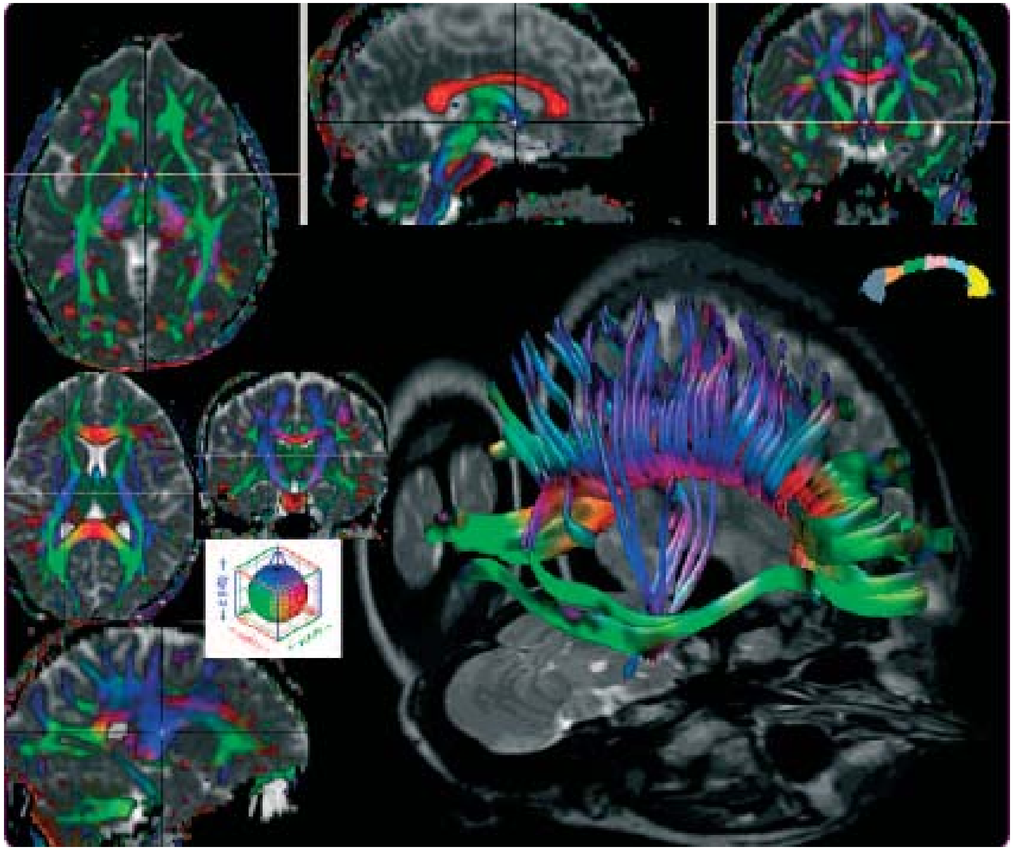
Fluid Attenuated
Inversion Recovery DTI

(a) proton density map, (b) T2w turbo spin echo map, (c) FLAIR map, (d) tissue segmentation map (white matter is white, gray matter is gray and CSF is cyan). Tensor decoding of the reference map (e) and the diffusion weighted images (f) with fusion of the DTI data (mean diffusivity map (h) fused with the fractional anisotropy map (i) modulated by the principal vector e1 (j)) results in the composite map (g). Further fusion of (g) and the tissue segmentation map (d) provides the map in (k).



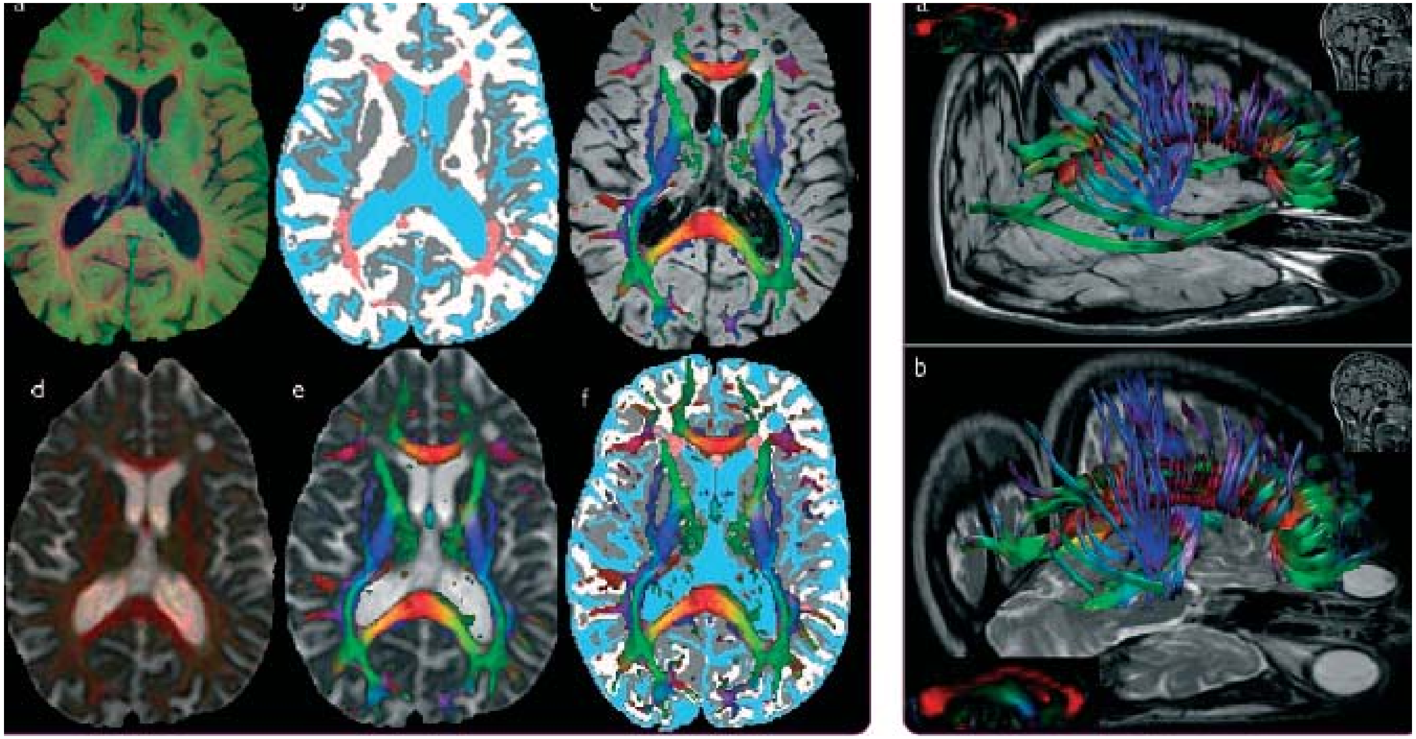
Fibers and Pathways

Connections of the callosal fibers (red:commissural fibers right-to-left) and the cortico-spinal track (blue: and association pathways (green: anterior-posterior).



Courtesy P.A. Narayan

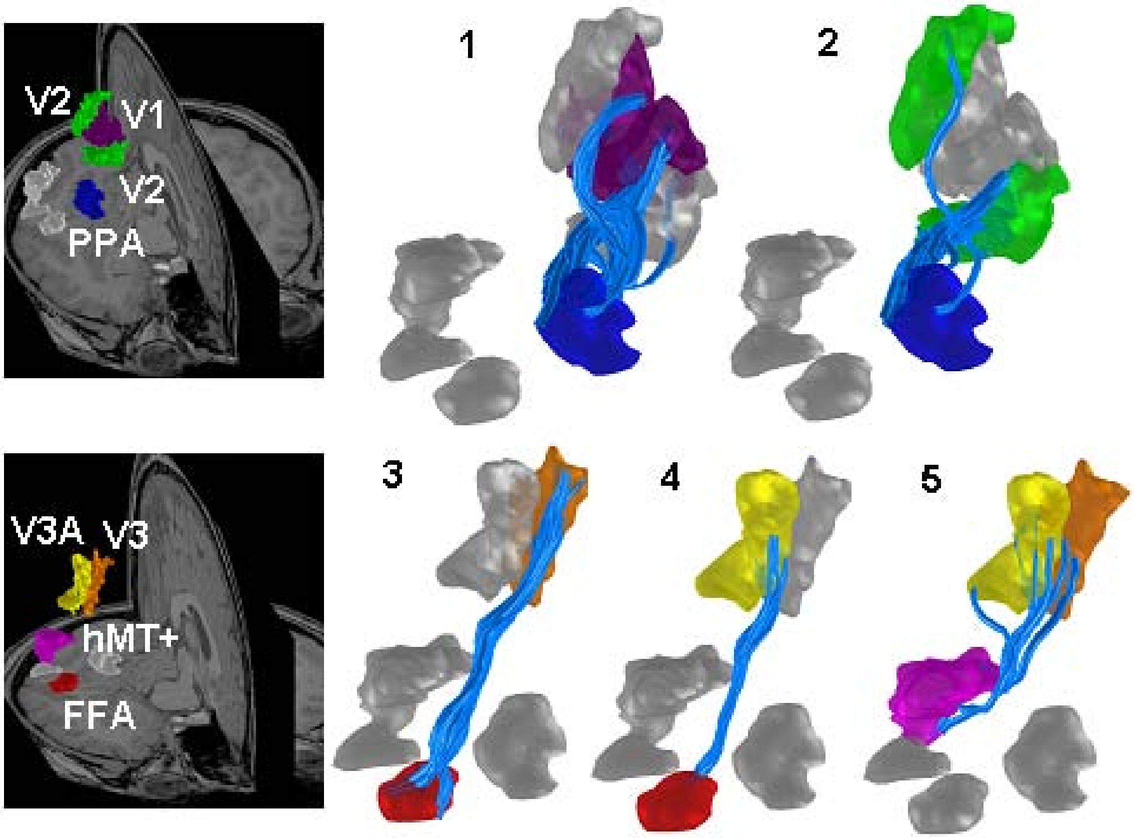
MS Case



(a) RGB fusion (FLAIR, phase sensitive inversion recovery ps-T1IR, post Gadolinium), (b) Conventional MRI tissue segmentation (PD,T2w, FLAIR), (c) |*FA*e1*| over FLAIR (d) RGB (DTI Eigenvalue Map) (e) |FA*e1| over mean diffusivity Dav (f) |FA*e1| segmented map in (b).

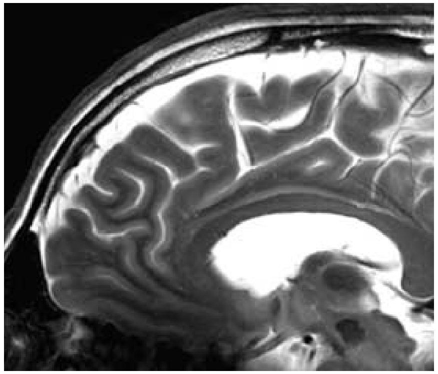
Loss of connectivity in the vicinity of the frontal lesion and the sustained tractability of the posterior callosal areas indicating possibly lesion activity, severity and duration.

Combining DTI fiber tractography with conventional fMRI



- High functional MRI (fMRI) activity during visual stimulation along the human ventro-temporal cortex are used as seeding points for DTI based fiber reconstructions.

7-T Imaging



The high signal-to-noise ratio available at 7.0T enables excellent spatial resolution.
T1-weighted 3D TFE with TR 19 ms,TE 9.5 ms, slices 1 mm, FOV 240 mm, matrix 700.

MS lesions can be seen in gray matter as well as white matter.

Courtesy Phillips Medical Systems