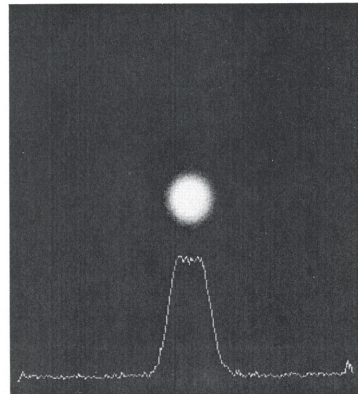
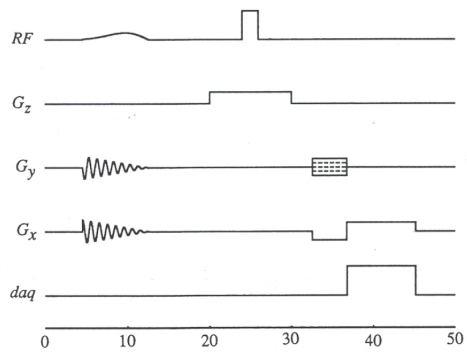


Bioengineering 280A Principles of Biomedical Imaging

Fall Quarter 2005
MRI Lecture 6

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Excitation k-space



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Pauly et al 1989

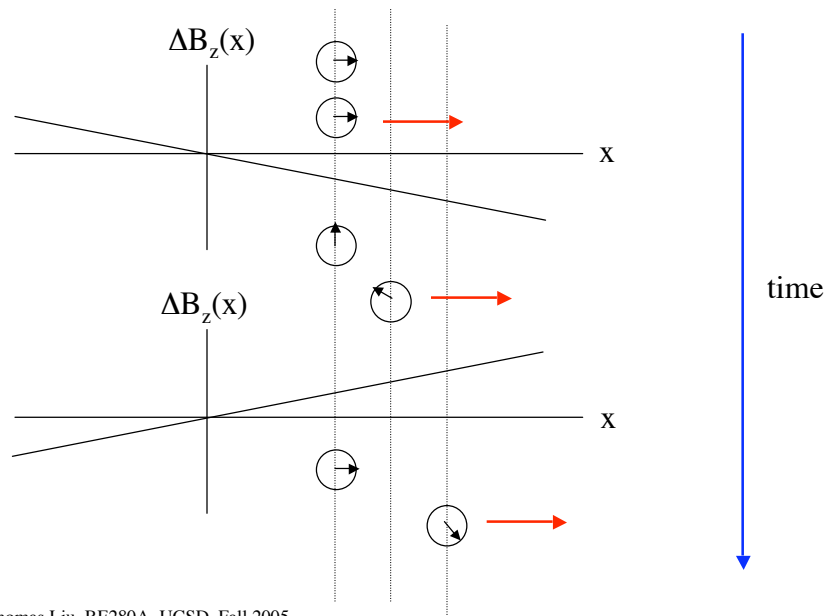
Moving Spins

So far we have assumed that the spins are not moving (aside from thermal motion giving rise to relaxation), and contrast has been based upon T_1 , T_2 , and proton density. We were able to achieve different contrasts by adjusting the appropriate pulse sequence parameters.

Biological samples are filled with moving spins, and we can also use MRI to image the movement. Examples: blood flow, diffusion of water in the white matter tracts. In addition, we can also sometimes induce motion into the object to image its mechanical properties, e.g. imaging of stress and strain with MR elastography.

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Phase of Moving Spin



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Phase of a Moving Spin

$$\begin{aligned}\varphi(t) &= -\int_0^t \Delta\omega(\tau) d\tau \\ &= -\int_0^t \gamma \Delta B(\tau) d\tau \\ &= -\int_0^t \gamma \vec{G}(\tau) \cdot \vec{r}(\tau) d\tau \\ &= -\gamma \int_0^t [G_x(\tau)x(\tau) + G_y(\tau)y(\tau) + G_z(\tau)z(\tau)] d\tau\end{aligned}$$

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Phase of Moving Spin

Consider motion along the x-axis

$$x(t) = x_0 + vt + \frac{1}{2}at^2$$

$$\begin{aligned}\varphi(t) &= -\gamma \int_0^t G_x(\tau)x(\tau) d\tau \\ &= -\gamma \int_0^t G_x(\tau) \left[x_0 + v\tau + \frac{1}{2}a\tau^2 \right] d\tau \\ &= -\gamma \left[x_0 \int_0^t G_x(\tau) d\tau + v \int_0^t G_x(\tau) \tau d\tau + \frac{a}{2} \int_0^t G_x(\tau) \tau^2 d\tau \right] \\ &= -\gamma \left[x_0 M_0 + v M_1 + \frac{a}{2} M_2 \right]\end{aligned}$$

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Phase of Moving Spin

$$\varphi(t) = -\gamma \left[x_0 M_0 + v M_1 + \frac{a}{2} M_2 \right]$$

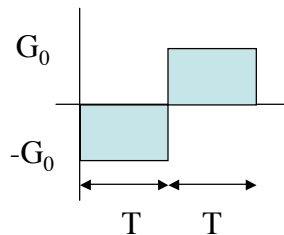
$$M_0 = \int_0^t G_x(\tau) d\tau \quad \text{Zeroth order moment}$$

$$M_1 = \int_0^t G_x(\tau) \tau d\tau \quad \text{First order moment}$$

$$M_2 = \int_0^t G_x(\tau) \tau^2 d\tau \quad \text{Second order moment}$$

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Flow Moment Example



$$M_0 = \int_0^t G_x(\tau) d\tau = 0$$

$$M_1 = \int_0^t G_x(\tau) \tau d\tau$$

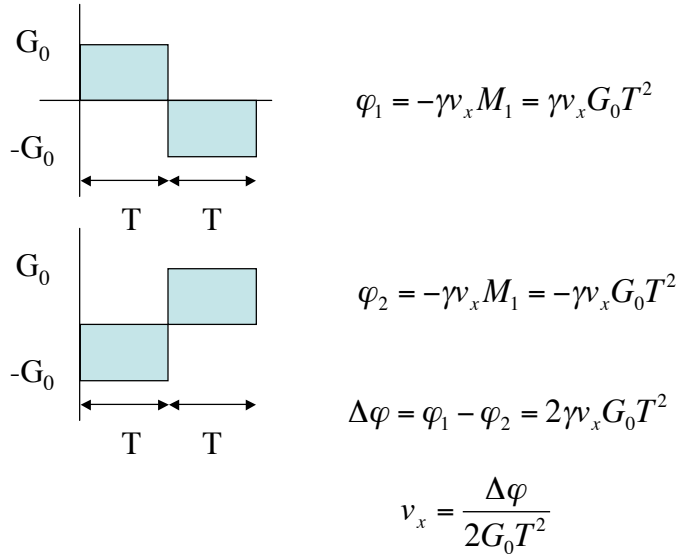
$$= -\int_0^T G_0 \tau d\tau + \int_T^{2T} G_0 \tau d\tau$$

$$= G_0 \left[-\frac{\tau^2}{2} \Big|_0^T + \frac{\tau^2}{2} \Big|_T^{2T} \right]$$

$$= G_0 \left[-\frac{T^2}{2} + \frac{4T^2}{2} - \frac{T^2}{2} \right] = G_0 T^2$$

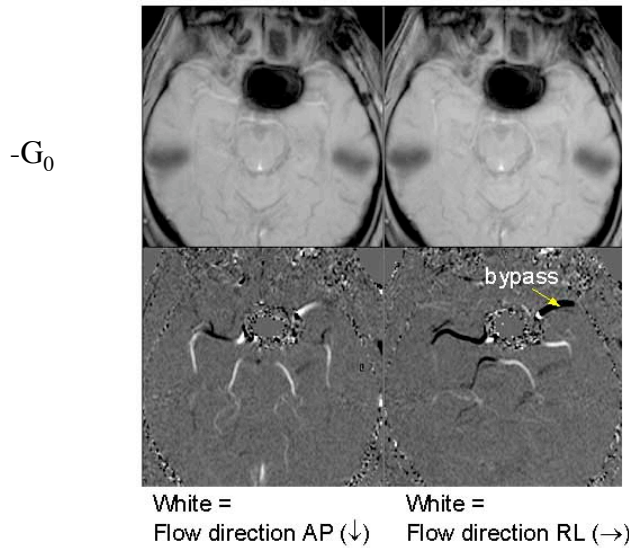
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Phase Contrast Angiography (PCA)



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PCA example



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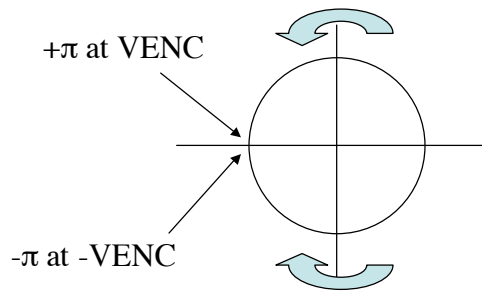
http://www.medical.philips.com/main/products/mri/assets/images/case_of_week/cotw_51_s5.jpg

Aliasing in PCA

Define VENC as the velocity at which the phase is 180 degrees.

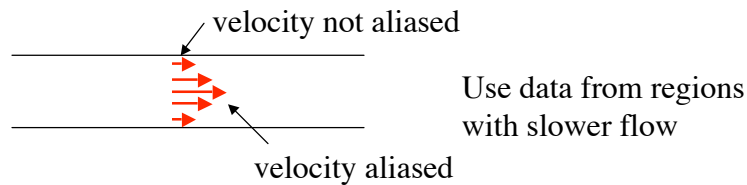
$$VENC \equiv \frac{\pi}{\gamma G_0 T^2}$$

Because of phase wrapping the velocity of spins flowing faster than VENC is ambiguous.



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Aliasing Solutions



Use multiple VENC values so that the phase differences are smaller than π radians.

$$\varphi_1 = \pi \frac{v_x}{VENC_1}$$

$$\varphi_2 = \pi \frac{v_x}{VENC_2}$$

$$\varphi_1 - \varphi_2 = \pi v_x \left(\frac{1}{VENC_1} - \frac{1}{VENC_2} \right)$$

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Velocity k-space

A bipolar gradient introduces a phase modulation across velocities of the form $\varphi(v_x) = -\gamma v_x G_0 T^2$

We can make measurements with different amounts of phase modulation and then integrate over velocities to obtain

$$\begin{aligned} M(k_{v_x}) &= \int_{-\infty}^{\infty} m(v_x) e^{j\varphi(v_x)} dv_x \\ &= \int_{-\infty}^{\infty} m(v_x) e^{-j\gamma v_x G_0 T^2} dv_x \\ &= \int_{-\infty}^{\infty} m(v_x) e^{-j2\pi k_{v_x} v_x} dv_x \\ &= F[m(v_x)] \quad \text{with } k_{v_x} = \frac{\gamma}{2\pi} G_0 T^2 \end{aligned}$$

By making measurements with bipolar gradients of varying amplitudes/durations and taking the inverse transform of the measurements, we can obtain the velocity distribution.

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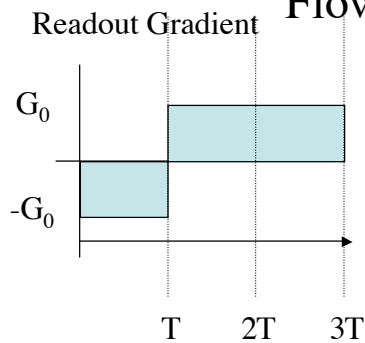
Velocity k-space

$$M(k_x, k_{v_x}) = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} m(x, v_x) e^{-j2\pi k_x x} e^{-j2\pi k_{v_x} v_x} dx dv_x$$

In addition, we can apply imaging gradients so that we can eventually obtain the velocity distribution at each point in space. A full k-space acquisition would then yield 6 dimensions -- 3 spatial dimensions and 3 velocity dimensions.

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Flow Artifacts



During readout moving spins within the object will accumulate phase that is in addition to the phase used for imaging. This leads to

- 1) Net phase at echo time $TE = 2T$.
- 2) An apparent shift in position of the object.
- 3) Blurring of the object due to a quadratic phase term.

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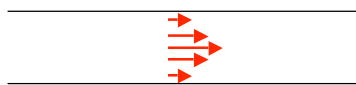
Flow Artifacts

Plug Flow



All moving spins in the voxel experience the same phase shift at echo time.

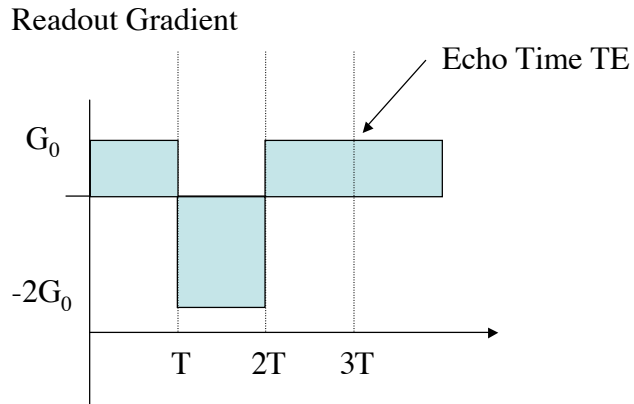
Laminar Flow



Spins have different phase shifts at echo time. The dephasing causes the cancellation and signal dropout.

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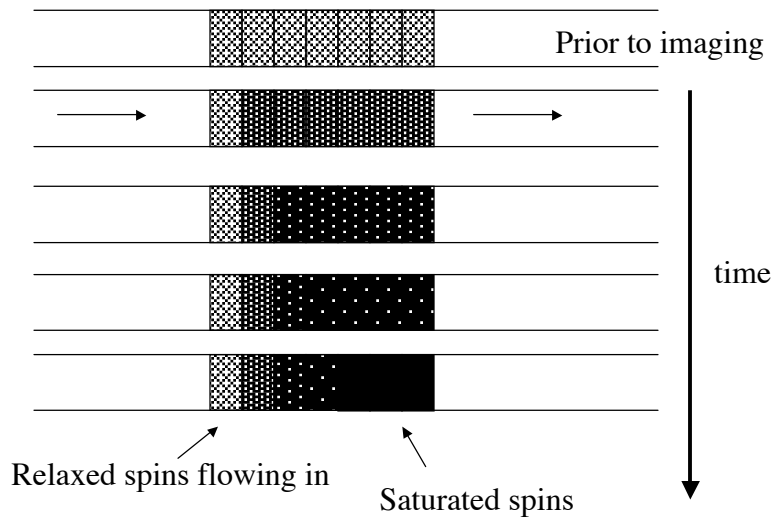
Flow Compensation



At TE both the first and second order moments are zero, so both stationary and moving spins have zero net phase.

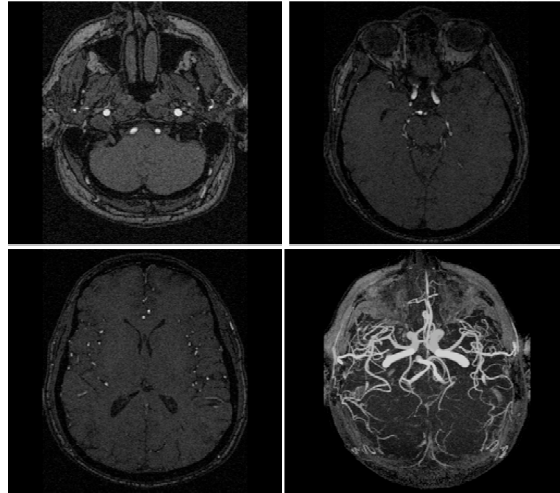
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Inflow Effect



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Time of Flight Angiography



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Cerebral Blood Flow (CBF)

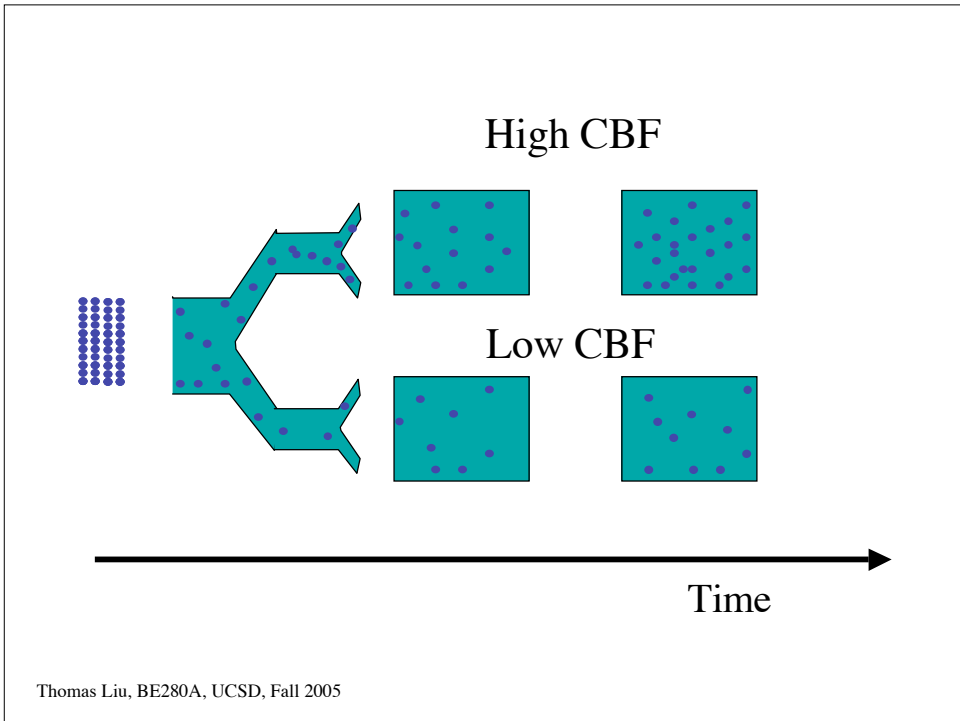
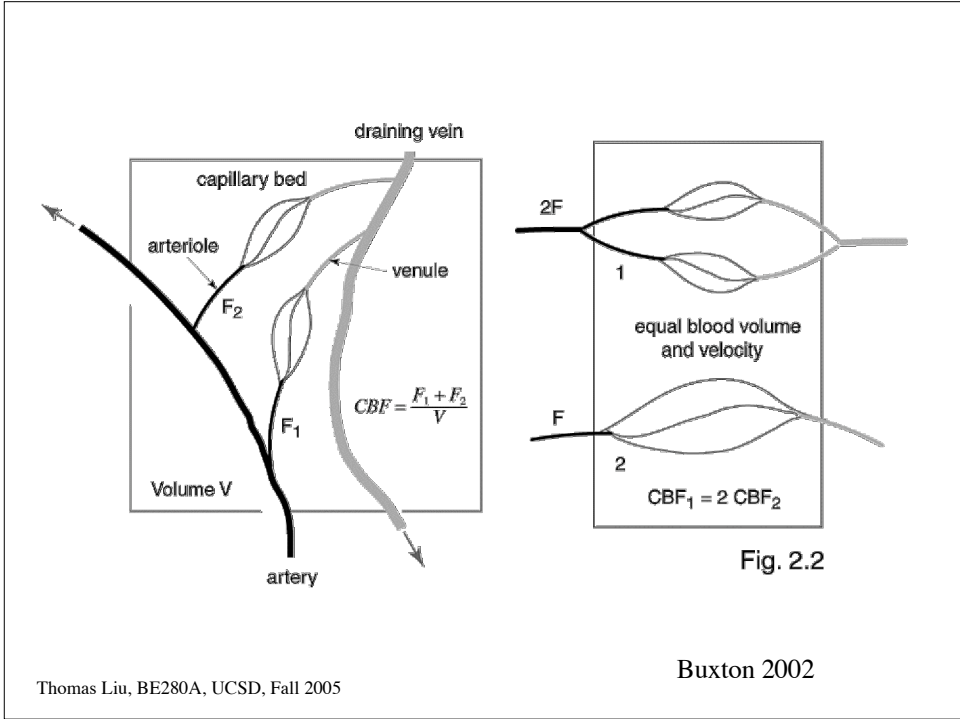
CBF = Perfusion

= Rate of delivery of arterial blood to a capillary bed in tissue.

Units: $\frac{\text{(ml of Blood)}}{\text{(100 grams of tissue)(minute)}}$

Typical value is 60 ml/(100g-min) or
 $60 \text{ ml}/(100 \text{ ml-min}) = 0.01 \text{ s}^{-1}$, assuming
average density of brain equals 1 gm/ml

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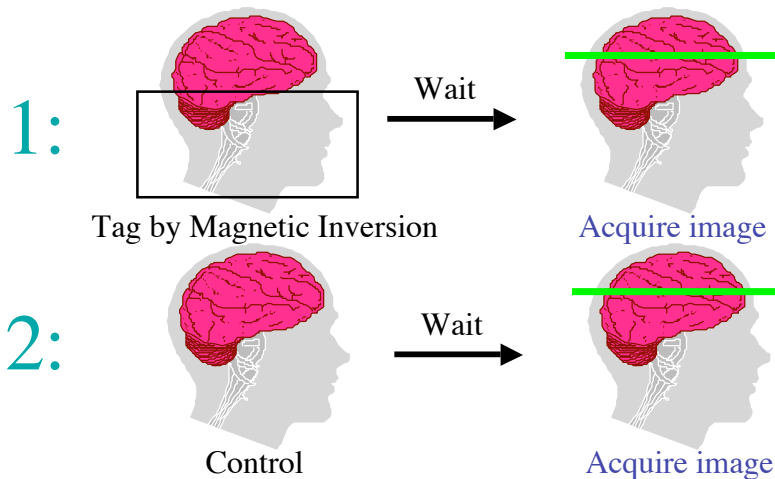


Arterial Spin Labeling

- Magnetically tag inflowing arterial blood
- Wait for tagged blood to flow into imaging slice
- Acquire image of tissue+tagged blood
- Apply control pulse that doesn't tag blood
- Acquire control image of tissue
- Control image-tag image = blood image

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Arterial Spin Labeling (ASL)



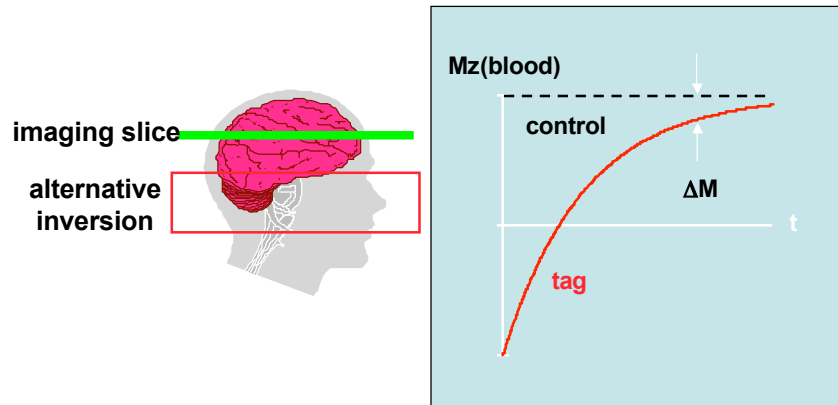
$$\text{Control} - \text{Tag} \propto \text{CBF}$$

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Credit: Wen-Ming Luh

Arterial Spin Labeling (ASL)

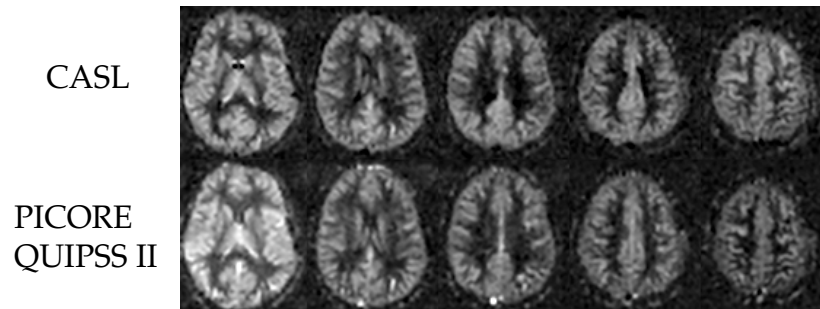
- water protons as freely diffusible tracers



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Courtesy of Wen-Ming Luh

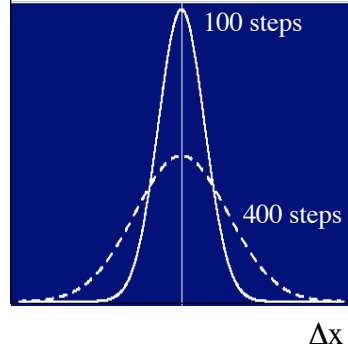
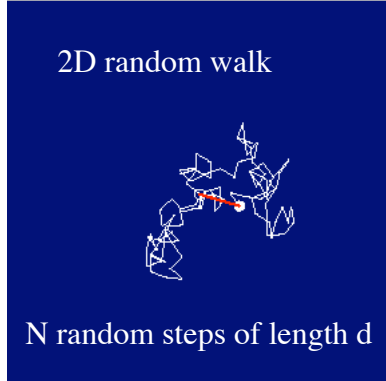
Multislice CASL and PICORE



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Credit: E. Wong

Diffusion



$$\langle \Delta x^2 \rangle = Nd^2 = 2DT$$

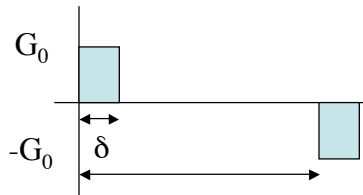
D = diffusivity

In brain:
 $D \approx 0.001 \text{ mm}^2/\text{s}$
 For $T=100 \text{ msec}$,
 $\Delta x \approx 15 \mu$

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Credit: Larry Frank

Diffusion Weighting



Assume $\delta \ll T$

$$\varphi(t_1) \approx -\gamma G_0 x(t_1) \delta$$

$$\varphi(t_2) \approx +\gamma G_0 x(t_2) \delta$$

Net Phase T

$$\varphi \approx \varphi(t_1) + \varphi(t_2) = \gamma G_0 [x(t_2) - x(t_1)] \delta = \gamma G_0 \Delta x \delta$$

Average Squared Phase

$$\langle \varphi^2 \rangle = \gamma^2 G_0^2 \delta^2 \langle (\Delta x)^2 \rangle = \gamma^2 G_0^2 \delta^2 2DT$$

Signal

$$S \propto e^{-\langle \varphi^2 \rangle / 2} = e^{-\gamma^2 G_0^2 \delta^2 DT} = e^{-bD} \text{ where } b = \gamma^2 G_0^2 \delta^2 T$$

$$\text{A more careful analysis yields } b = \gamma^2 G_0^2 \delta^2 (T - \delta/3)$$

b-factor

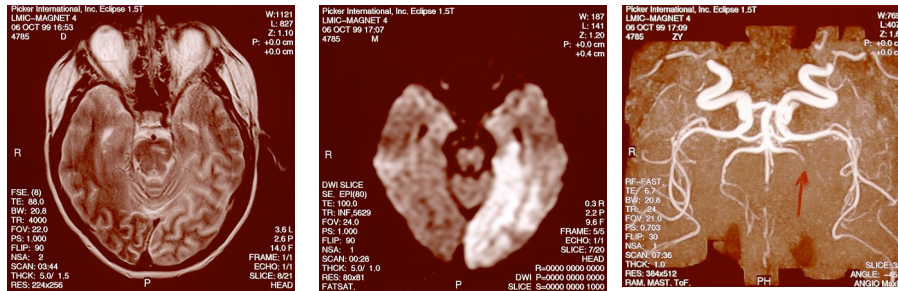
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Diffusion Weighted Images

T2 weighted

Diffusion Weighted

Angiogram



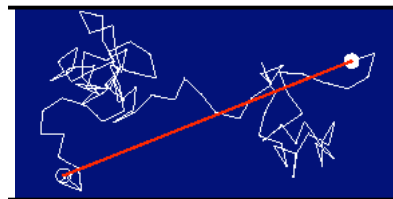
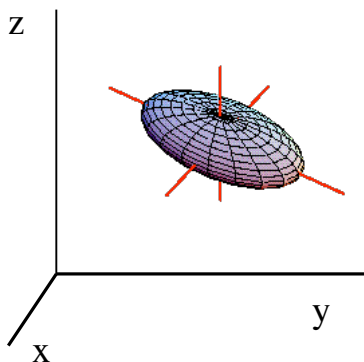
After a stroke, normal water movement is restricted in the region of damage. Diffusivity decreases, so the signal intensity increases.

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<http://lehighmri.com/cases/dwi/patient-b.html>

Restricted Diffusion

D depends on direction

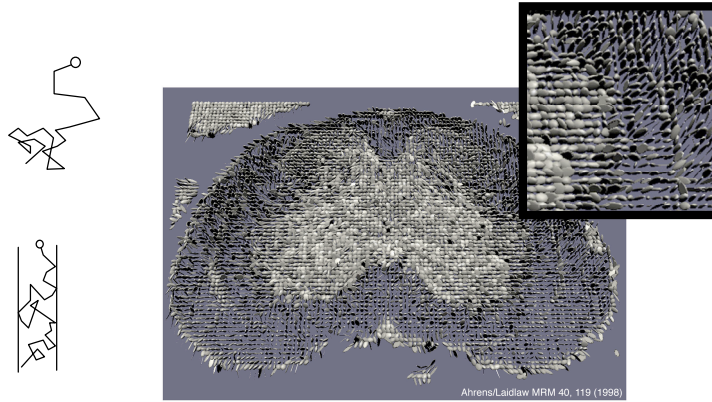


Diffusion tensor:
3 values of D
3 angles

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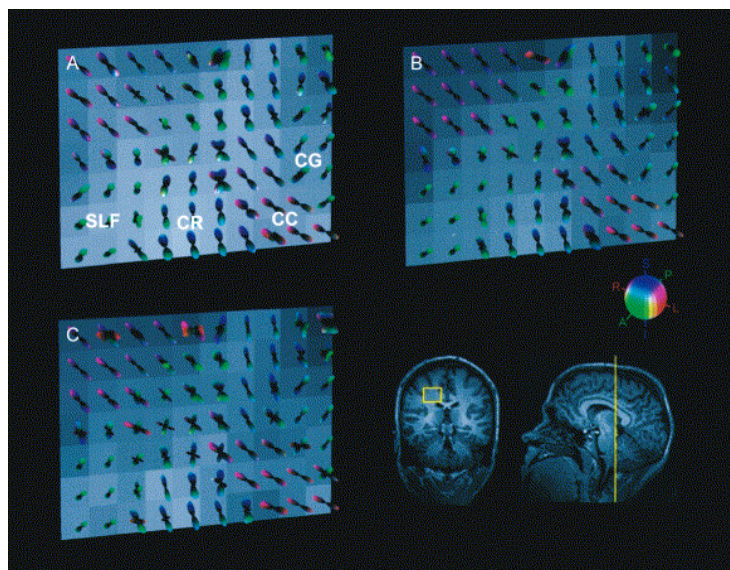
Credit: Larry Frank

Diffusion Imaging Example



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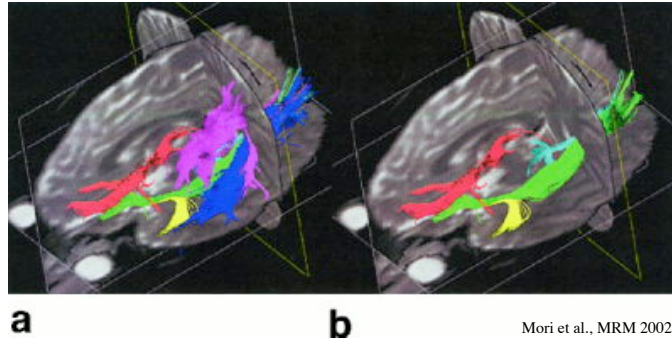
Q-ball imaging



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Tuch et al, Neuron 2003

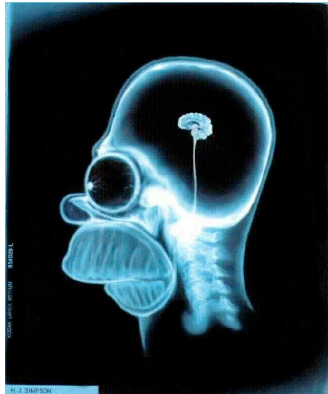
Fiber Tract Mapping



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fMRI

MRI studies brain anatomy.



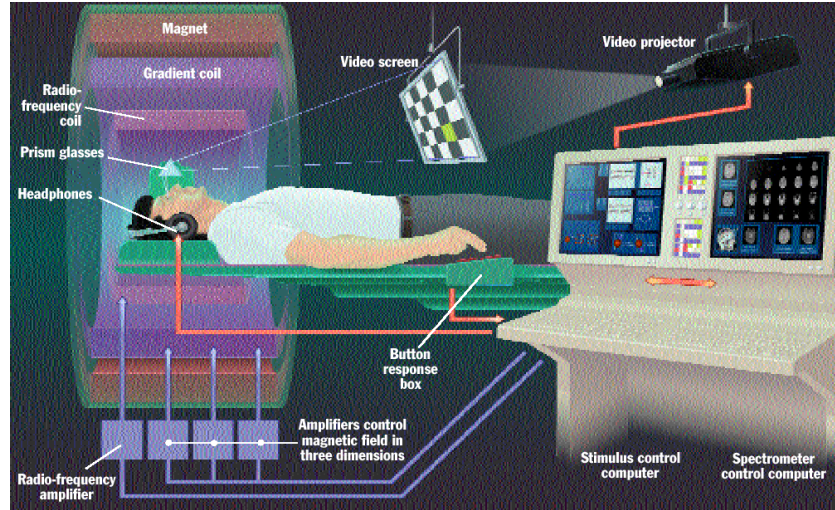
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**Functional MRI (fMRI)
studies brain function.**



http://defiant.ssc.uwo.ca/lody_web/fmri4dummies.htm

fMRI Setup



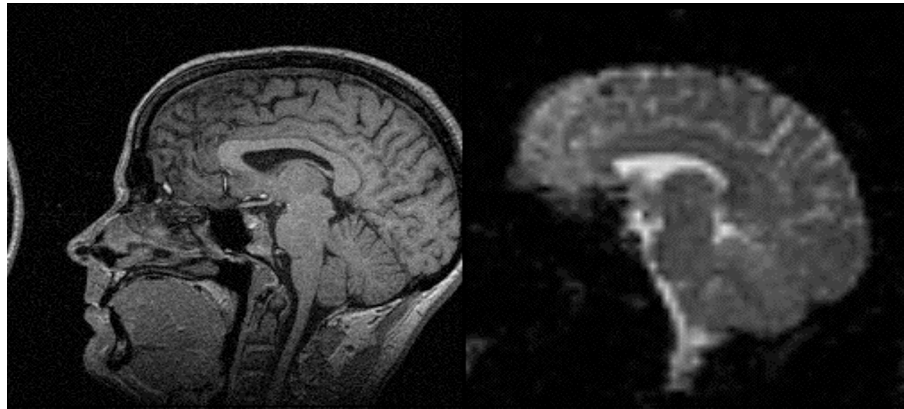
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http://defiant.ssc.uwo.ca/Iody_web/fmri4dummies.htm

fMRI Acquisition

High spatial resolution

High temporal resolution



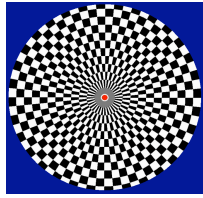
MP-RAGE
Voxel volume: 1 mm^3
Imaging time: 6 min

EPI
Voxel volume: 45 mm^3
Imaging time: 60 msec

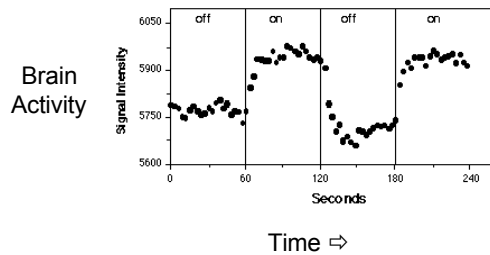
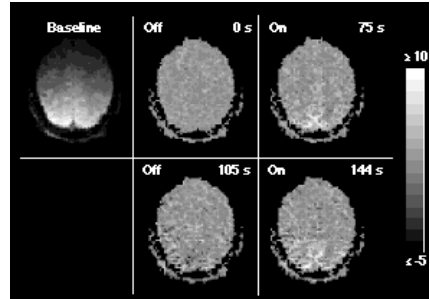
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Buxton 2002

Visual Activation



Flickering Checkerboard
OFF (60 s) - ON (60 s) - OFF (60 s) - ON (60 s) - OFF (60 s)

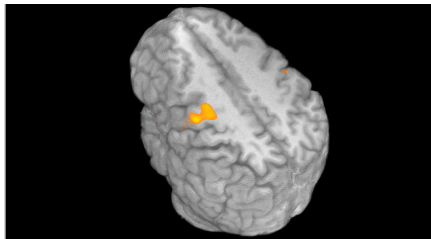


Source: Kwong et al., 1992

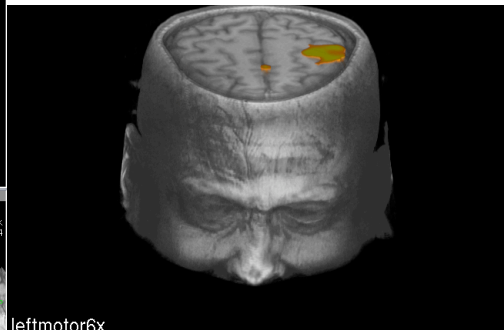
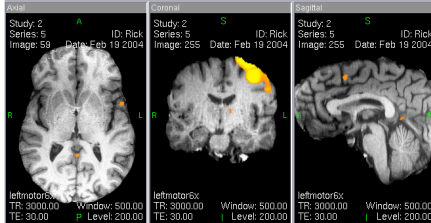
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http://defiant.ssc.uwo.ca/Jody_web/fmri4dummies.htm

Finger Tapping Task



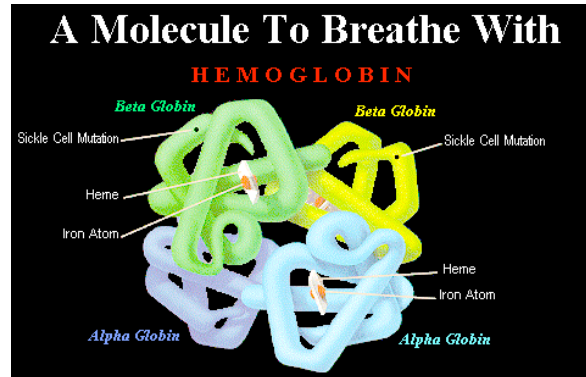
leftmotor6x



leftmotor6x

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Hemoglobin



Oxygen binds to the iron atoms to form oxyhemoglobin HbO_2
Release of O_2 to tissue results in deoxyhemoglobin dHbO_2

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<http://www.people.virginia.edu/~rjh9u/hemoglob.html>

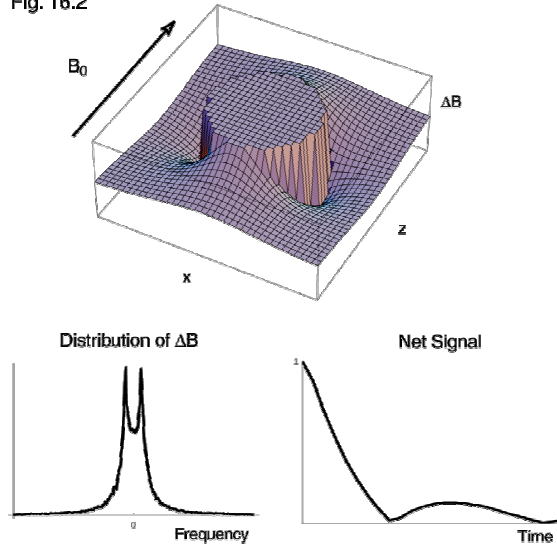
Effect of dHbO_2

dHbO_2 is paramagnetic due to the iron atoms. As it becomes oxygenated, it becomes less paramagnetic.

dHbO_2 perturbs the local magnetic fields. As blood becomes more deoxygenated, the amount of perturbation increases and there is more dephasing of the spins. Thus as dHbO_2 increases we find that T_2^* decreases and the amplitude $\exp(-TE/T_2^*)$ image of a T_2^* weighted image will decrease. Conversely as dHbO_2 decreases, T_2^* increases and we expect the signal amplitude to go up.

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Field Distortions Around a Magnetized Blood Vessel
Fig. 16.2



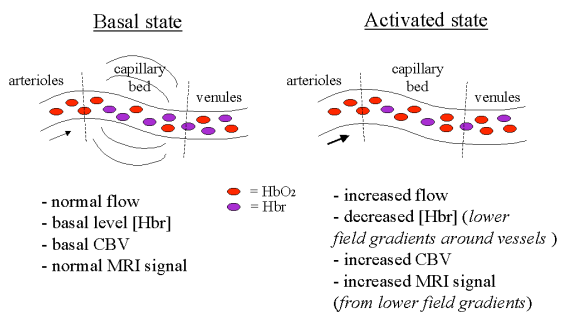
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Buxton 2002

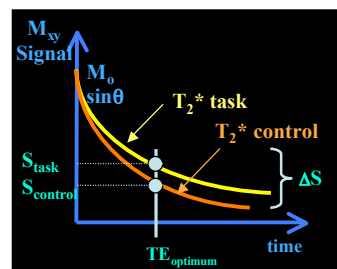
BOLD Effect

Blood Oxygen Level Dependent signal

neural activity \rightarrow \uparrow blood flow \rightarrow \uparrow oxyhemoglobin \rightarrow \uparrow T_2^ \rightarrow \uparrow MR signal



Source: [fMRIB Brief Introduction to fMRI](#)

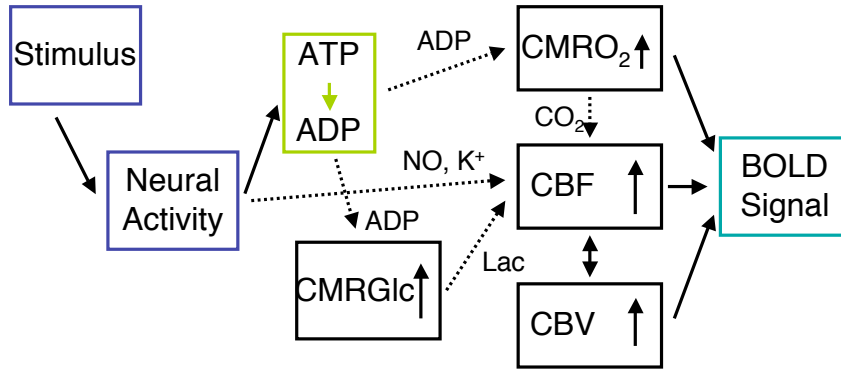


Source: Jorge Jovicich

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http://defiant.ssc.uwo.ca/Jody_web/fmri4dummies.htm

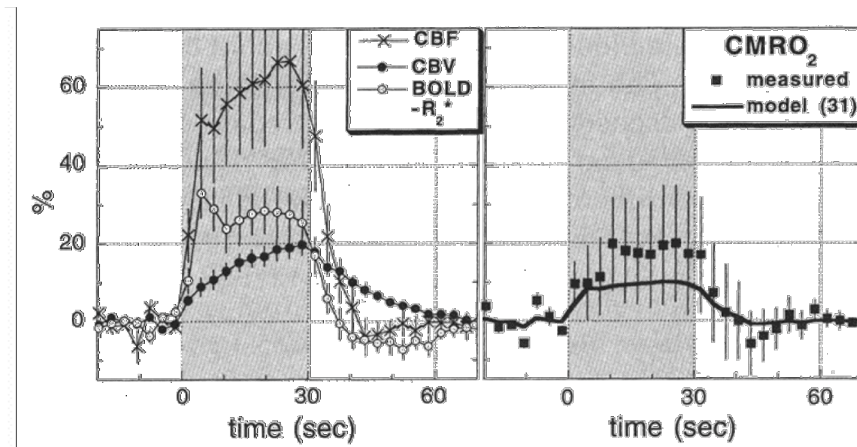
BOLD Dynamics



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Credit: Rick Buxton

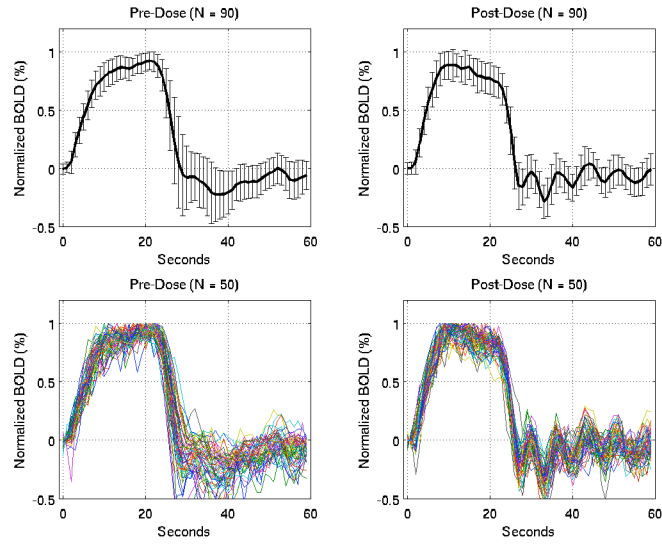
Fig. 17.5
BOLD Dynamics
Rat Forepaw Stimulation
 (Mandeville, et al, 1999)



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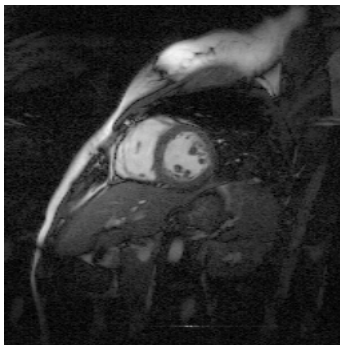
Buxton 2002

BOLD and Vascular Dynamics



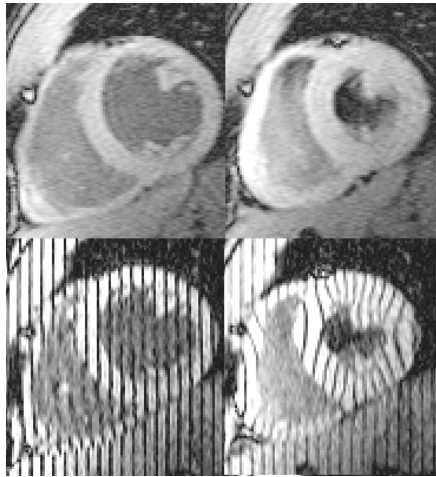
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Cardiac Imaging



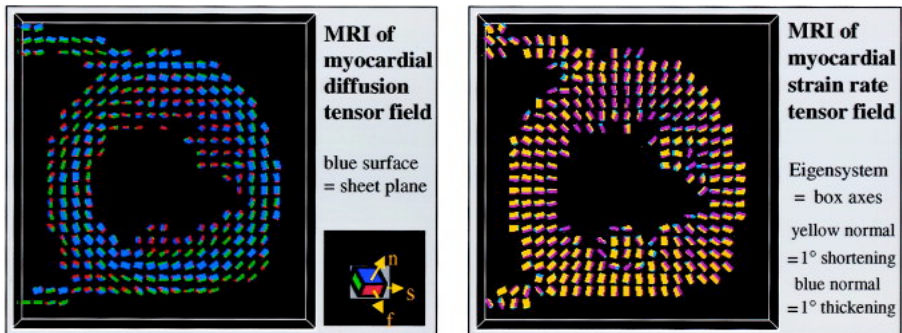
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Cardiac Tagging



Thomas Liu, BE280A, UCSD, Fall 2005

Cardiac Diffusion and Strain



Dou et al, MRM 2003

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