Bioengineering 280A
Principles of Biomedical Imaging

Fall Quarter 2004
MRI Lecture 4

Topics

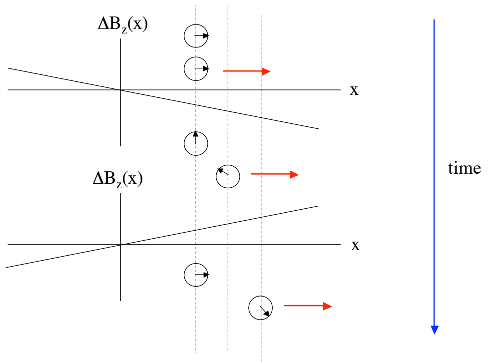
- Velocity Imaging
- Perfusion Imaging
- Diffusion Imaging
- Functional MRI

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.

Phase of Moving Spin



Phase of a Moving Spin

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

Phase of Moving Spin

Consider motion along the x-axis

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

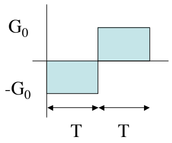
$$
\begin{aligned}
\varphi(t) &= -\gamma\int_0^t G_x(\tau)x(\tau)\,d\tau \\
&= -\gamma\int_0^t G_x(\tau)\left[x_0 + v\tau + \frac{1}{2}a\tau^2\right]d\tau \\
&= -\gamma\left[x_0\int_0^t G_x(\tau)\,d\tau + v\int_0^t G_x(\tau)\tau\,d\tau + \frac{a}{2}\int_0^t G_x(\tau)\tau^2\,d\tau\right] \\
&= -\gamma\left[x_0 M_0 + vM_1 + \frac{a}{2}M_2\right]
\end{aligned}
$$

Phase of Moving Spin

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

$M_0 = \int_0^t G_x(\tau)d\tau$	Zeroth order moment
$M_1 = \int_0^t G_x(\tau)\tau d\tau$	First order moment
$M_2 = \int_0^t G_x(\tau)\tau^2 d\tau$	Second order moment

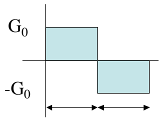
Flow Moment Example



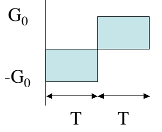
$$M_0 = \int_0^t G_x(\tau)d\tau = 0$$

$$\begin{aligned} M_1 &= \int_0^t G_x(\tau)\tau d\tau \\ &= -\int_0^T G_0 \tau d\tau + \int_T^{2T} G_0 \tau d\tau \\ &= G_0\left[-\left.\frac{\tau^2}{2}\right|_0^T + \left.\frac{\tau^2}{2}\right|_T^{2T}\right] \\ &= G_0\left[-\frac{T^2}{2} + \frac{4T^2}{2} - \frac{T^2}{2}\right] = G_0 T^2 \end{aligned}$$

Phase Contrast Angiography (PCA)



$$\varphi_1 = -\gamma v_x M_1 = \gamma v_x G_0 T^2$$



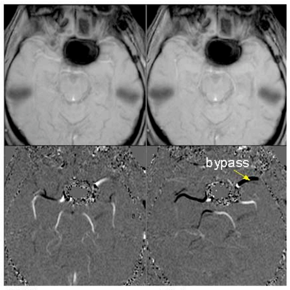
$$\varphi_2 = -\gamma v_x M_1 = -\gamma v_x G_0 T^2$$

$$\Delta\varphi = \varphi_1 - \varphi_2 = 2\gamma v_x G_0 T^2$$

$$v_x = \frac{\Delta\varphi}{2G_0 T^2}$$

PCA example

$-G_0$



White =
Flow direction AP (↓)

White =
Flow direction RL (→)

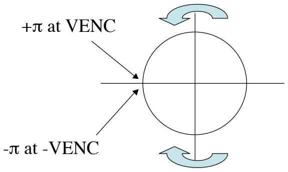
http://www.medical.philips.com/main/products/mri/assets/images/case_of_week/cotw_51_s5.jpg

Aliasing in PCA

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

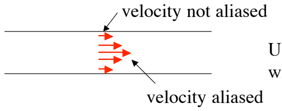
$$VENC \equiv \frac{\pi}{\gamma G_0 T^2}$$

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


+π at VENC

-π at -VENC

Aliasing Solutions


velocity not aliased

velocity aliased

Use data from regions with slower flow

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

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

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

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

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

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

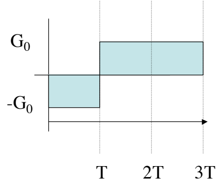
By making measurements with bipolar gradients of varying amplitudes/durations and taking the inverse transform of the measurements, we can obtain the velocity distribution.

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 acqusition would then yield 6 dimensions -- 3 spatial dimensions and 3 velocity dimensions.

Flow Artifacts

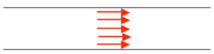


During readout moving spins within the object will accumulate phase that is in addition to the phase used for imaging. This leads to

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.

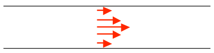
Flow Artifacts

Plug Flow



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

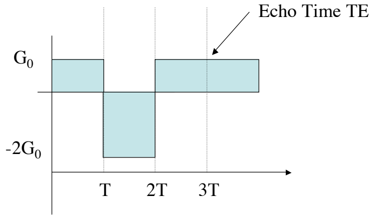
Laminar Flow



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

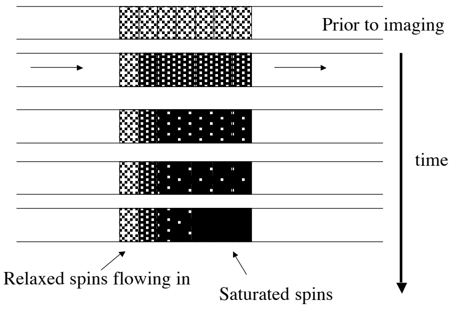
Flow Compensation

Readout Gradient

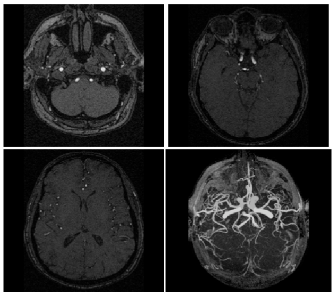


At TE both the first and second order moments are zero, so both stationary and moving spins have zero net phase.

Inflow Effect



Time of Flight Angiography



Cerebral Blood Flow (CBF)

CBF = Perfusion

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

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

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

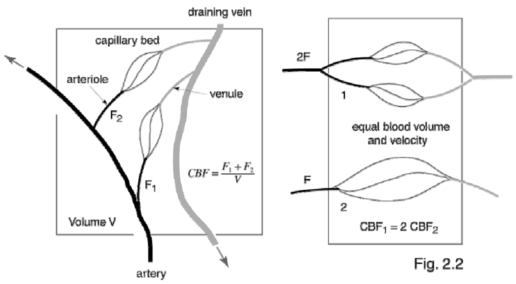
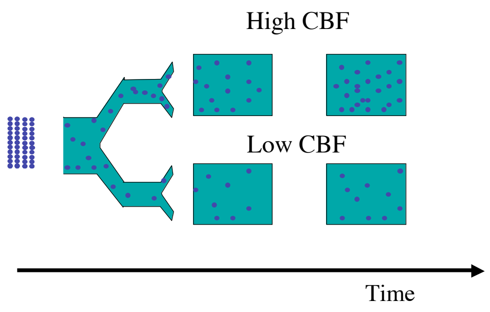


Fig. 2.2

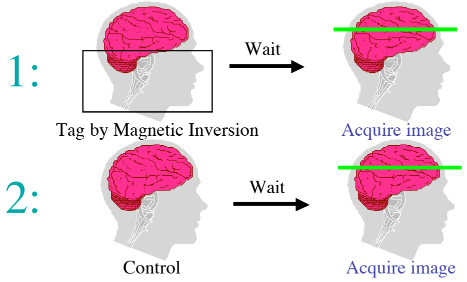
Buxton 2002



Arterial Spin Labeling

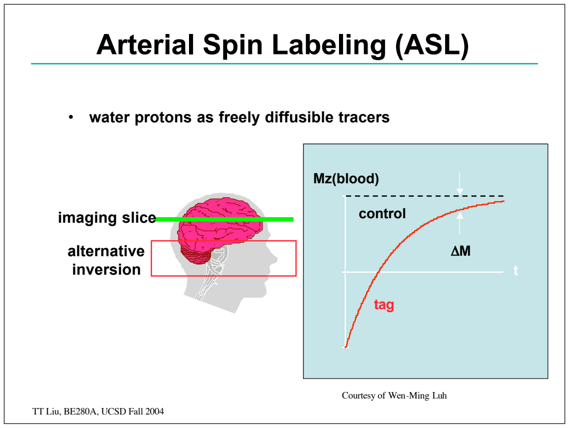
- Magnetically tag inflowing arterial blood
- Wait for tagged blood to flow into imaging slice
- Acquire image of tissue+tagged blood
- Apply control pulse that doesn't tag blood
- Acquire control image of tissue
- Control image-tag image = blood image

Arterial Spin Labeling (ASL)



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

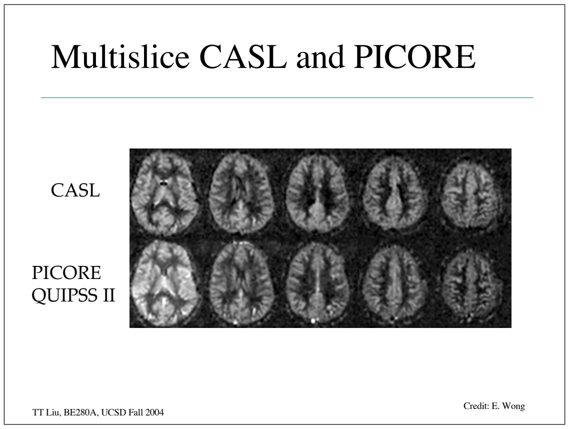
Credit: Wen-Ming Luh



Arterial Spin Labeling (ASL)

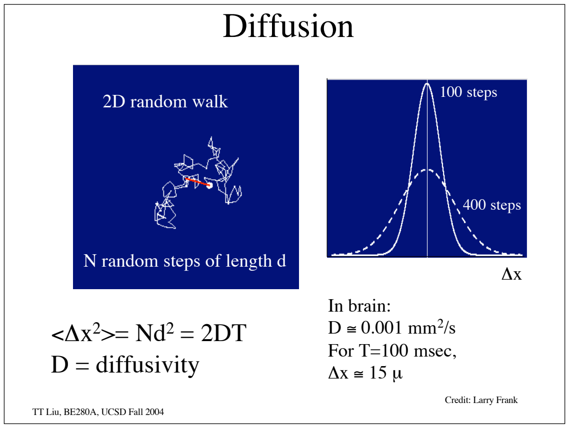
- water protons as freely diffusible tracers

Courtesy of Wen-Ming Luh



Multislice CASL and PICORE

Credit: E. Wong



Diffusion

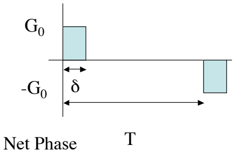
$<\Delta x^2> = Nd^2 = 2DT$

D = diffusivity

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

Credit: Larry Frank

Diffusion Weighting



Assume δ << 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

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

Average Squared Phase

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

Signal

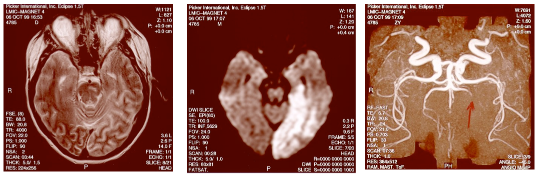
b-factor

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

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

Diffusion Weighted Images

T2 weighted Diffusion Weighted Angiogram

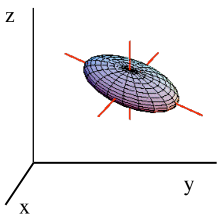


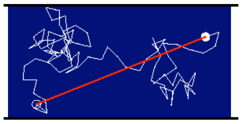
After a stroke, normal water movement is restricted in the region of damage. Diffusivity decreases, so the signal intensity increases.

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

Restricted Diffusion

D depends on direction

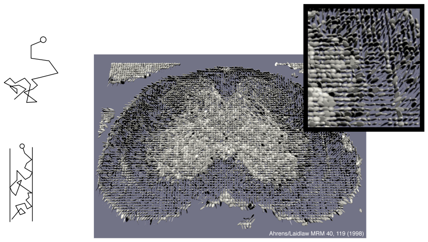




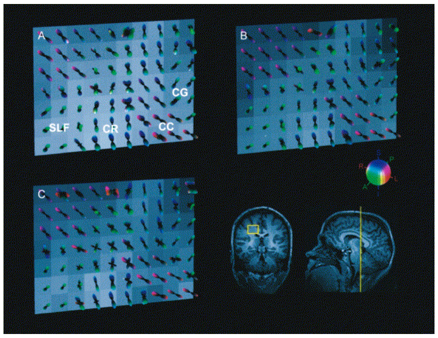
Diffusion tensor:
3 values of D
3 angles

Credit: Larry Frank

Diffusion Imaging Example

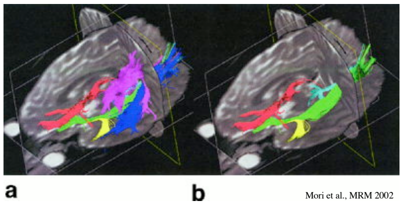

Ahrens/Laidlaw MRM 40, 119 (1998)

Q-ball imaging



Tuch et al, Neuron 2003

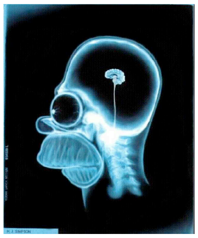
Fiber Tract Mapping



Mori et al., MRM 2002

fMRI

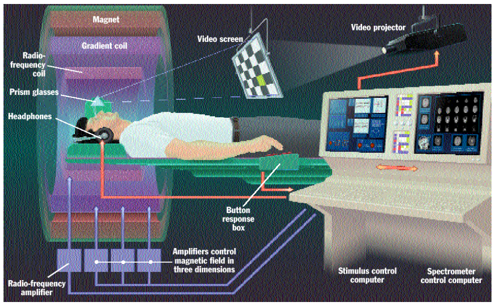
MRI studies brain anatomy.



Functional MRI (fMRI) studies brain function.

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

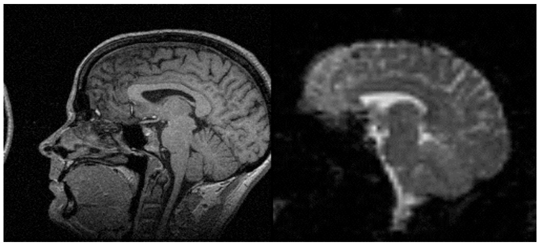
fMRI Setup



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

fMRI Acquisition

High spatial resolution | High temporal resolution

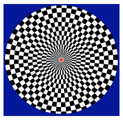


MP-RAGE
Voxel volume: 1 mm^3
Imaging time: 6 min

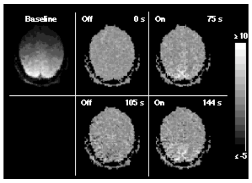
EPI
Voxel volume: 45 mm^3
Imaging time: 60 msec

Buxton 2002

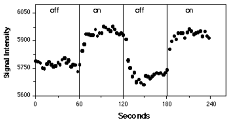
Visual Activation



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



Brain Activity



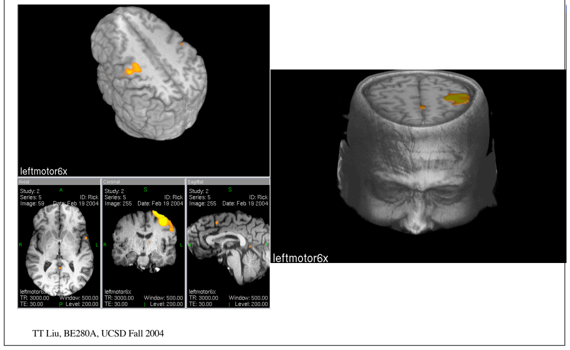
Time ⇨

Source: Kwong et al., 1992

TT Liu, BE280A, UCSD Fall 2004

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

Finger Tapping Task



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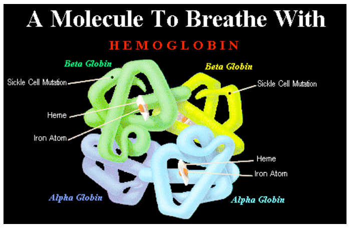
Hemoglobin



Oxygen binds to the iron atoms to form oxyhemoglobin HbO_2

Release of O_2 to tissue results in deoxyhemoglobin $dHBO_2$

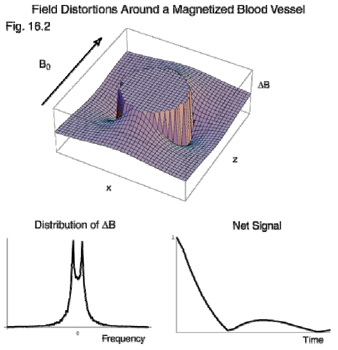
TT Liu, BE280A, UCSD Fall 2004

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.

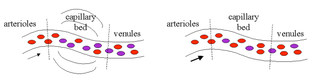


Buxton 2002

BOLD Effect

Blood Oxygen Level Dependent signal

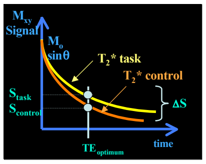
✸neural activity ➔ ↑ blood flow ➔ ↑ oxyhemoglobin ➔ ↑ T2* ➔ ↑ MR signal



- normal flow
- basal level [Hbr]
- basal CBV
- normal MRI signal

● = HbO_2
● = Hbr

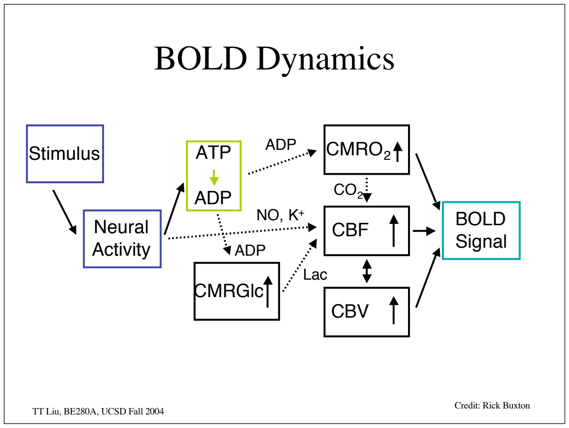
- increased flow
- decreased [Hbr] (*lower field gradients around vessels*)
- increased CBV
- increased MRI signal (*from lower field gradients*)



Source: fMRIB Brief Introduction to fMRI

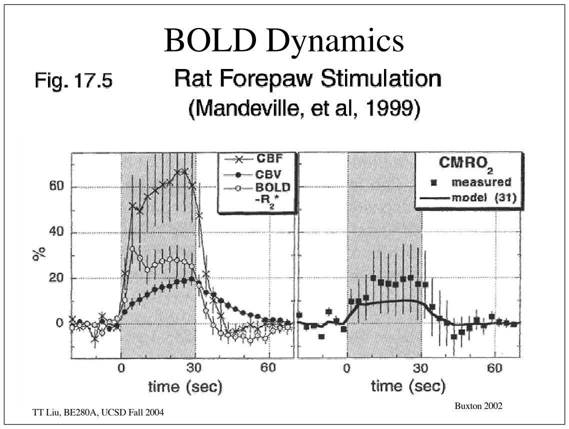
Source: Jorge Jovicich

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



BOLD Dynamics

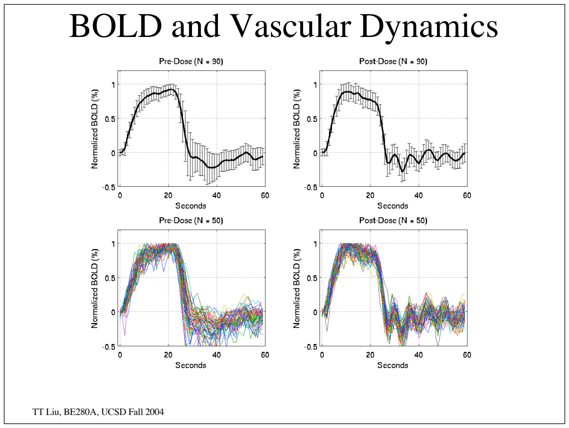
Credit: Rick Buxton



BOLD Dynamics

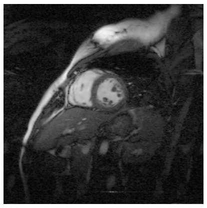
Fig. 17.5 Rat Forepaw Stimulation (Mandeville, et al, 1999)

Buxton 2002

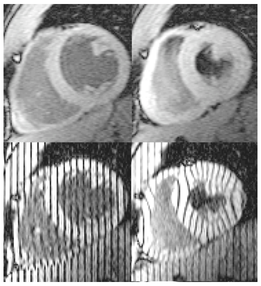


BOLD and Vascular Dynamics

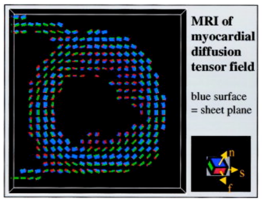
Cardiac Imaging

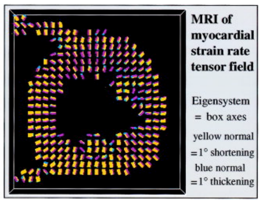


Cardiac Tagging



Cardiac Diffusion and Strain





Dou et al, MRM 2003