Bioengineering 280A
Principles of Biomedical Imaging

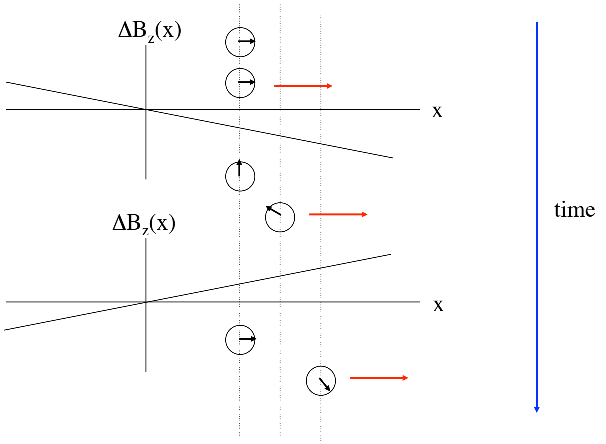
Fall Quarter 2009
MRI Lecture 6

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

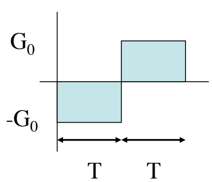
$$\varphi(t) = -\gamma\left[x_0 M_0 + vM_1 + \frac{a}{2}M_2\right]$$

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

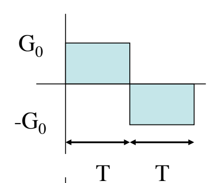
Flow Moment Example



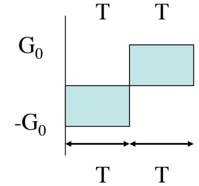
$$M_0 = \int_0^t G_x(\tau)d\tau = 0$$

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

Phase Contrast Angiography (PCA)



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



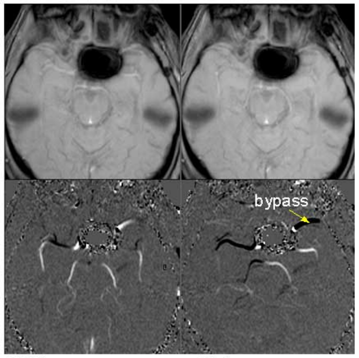
$$\varphi_2 = -\gamma v_x M_1 = -\gamma v_x G_0 T^2$$

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

$$v_x = \frac{\Delta\varphi}{2G_0T^2}$$

PCA example

-G_0



White =
Flow direction AP (↓)

White =
Flow direction RL (→)

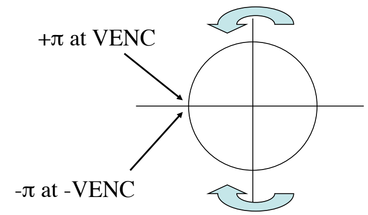
http://www.medical.philips.com/main/products/mri/assets/images/case_of_week/cotw_51_s5.jpg

Aliasing in PCA

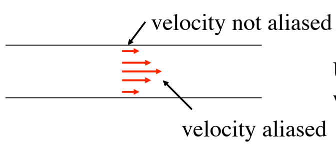
Define VENC as the velocity at at which the phase is 180 degrees.

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

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



Aliasing Solutions



Use data from regions with slower flow

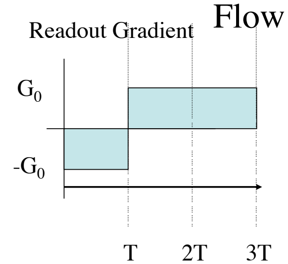
Use multiple VENC values so that the phase differences are smaller than π radians.

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

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

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

Flow Artifacts



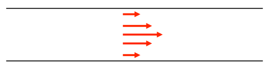
During readout moving spins within the object will accumulate phase that is in addition to the phase used for imaging. This leads to

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

Laminar Flow

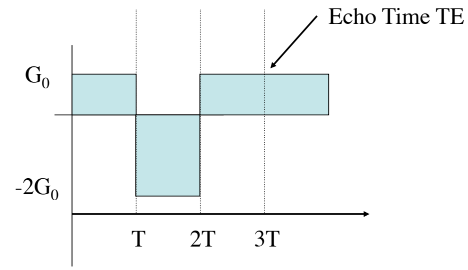


All moving spins in the voxel experience the same phase shift at echo time.

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

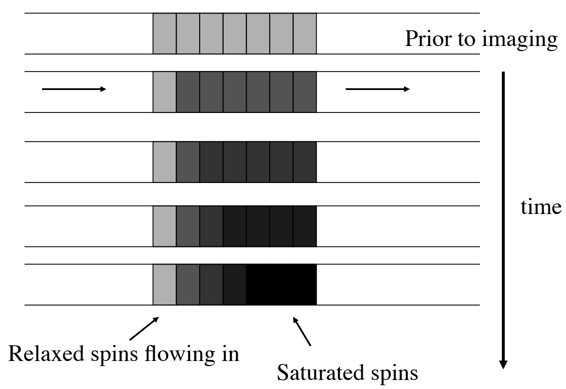
Flow Compensation

Readout Gradient

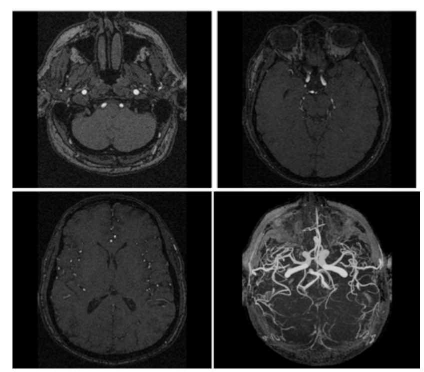


At TE both the first and second order moments are zero, so both stationary and moving spins have zero net phase.

Inflow Effect



Time of Flight Angiography

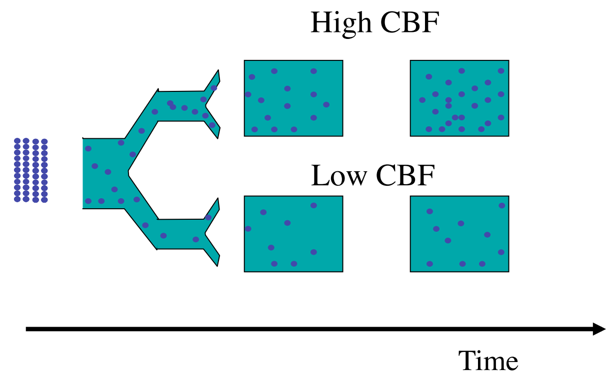


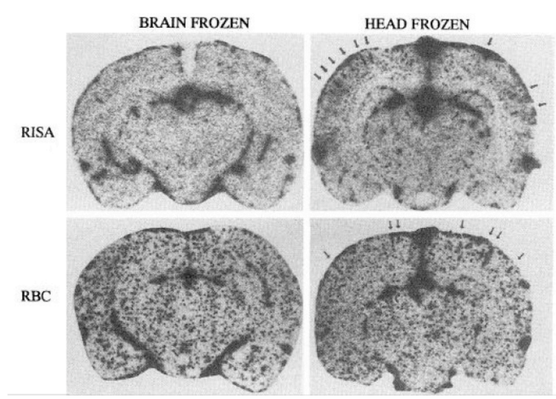
Cerebral Blood Flow (CBF)

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

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

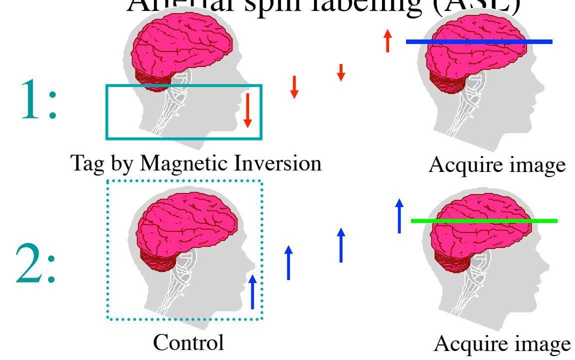
Typical value is 60 ml/(100g-min) or 60 ml/(100 ml-min) = 0.01 s^{-1}, assuming average density of brain equals 1 gm/ml





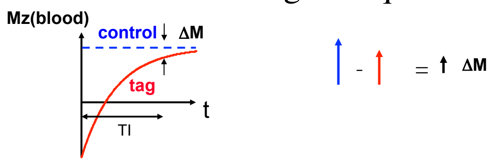
Bereczki et al 1992

Arterial spin labeling (ASL)



Control - Tag = $\Delta M \propto$ CBF

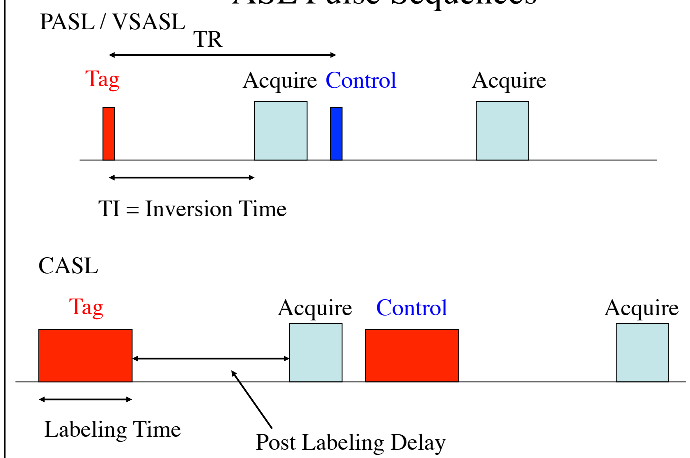
ASL Signal Equation



$$\Delta M = CBF \cdot A_{eff}$$

A_{eff} is the effective area of the arterial bolus. It depends on both physiology and pulse sequence parameters.

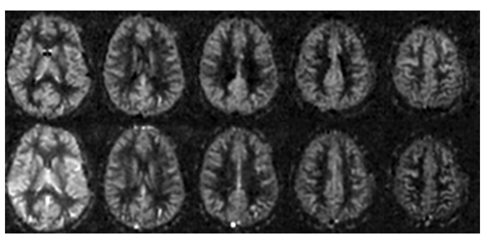
ASL Pulse Sequences



Multislice CASL and PICORE

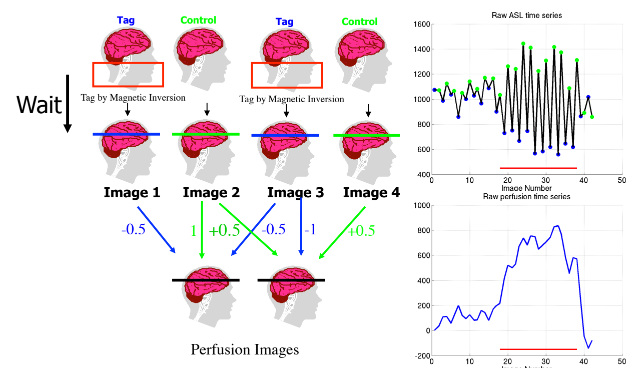
CASL

PICORE
QUIPSS II

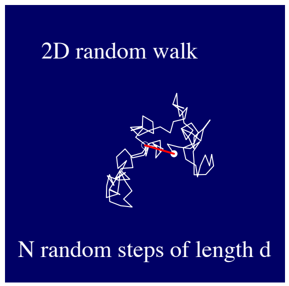


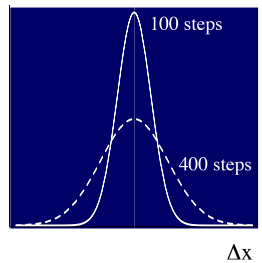
Credit: E. Wong

ASL Time Series



Diffusion





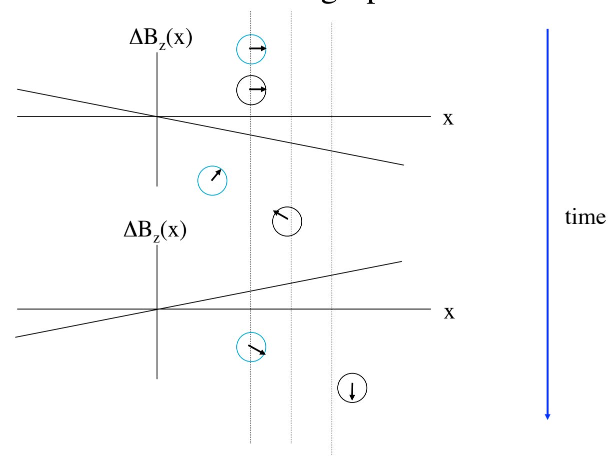
$<\Delta x^2> = Nd^2 = 2DT$

D = diffusivity

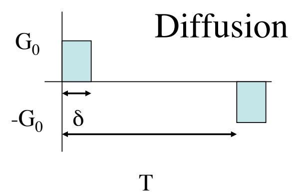
In brain:
$D \cong 0.001\ mm^2/s$
For T=100 msec,
$\Delta x \cong 15\ \mu$

Credit: Larry Frank

Diffusing Spins



Diffusion Weighting



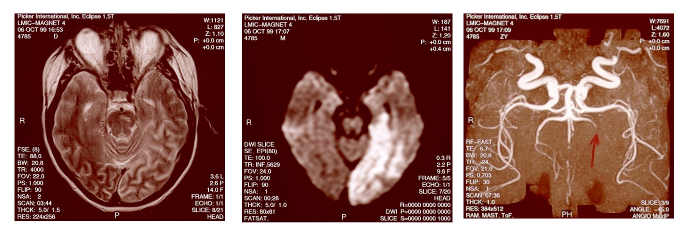
Signal

$$S \propto e^{-\gamma^2 G_0^2 \delta^2 DT} = e^{-bD} \quad \text{where } b = \gamma^2 G_0^2 \delta^2 (T - \delta/3)$$

Diffusivity

Diffusion Weighted Images

T2 weighted | Diffusion Weighted | Angiogram

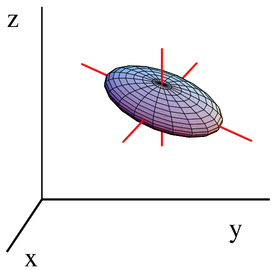


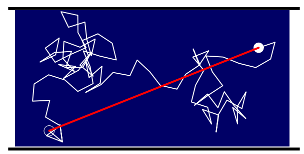
After a stroke, normal water movement is restricted in the region of damage. Diffusivity decreases, so the signal intensity increases.

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

Restricted Diffusion

D depends on direction

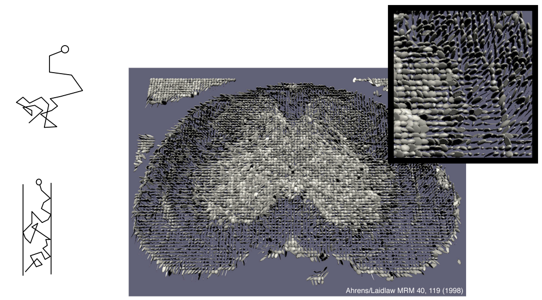




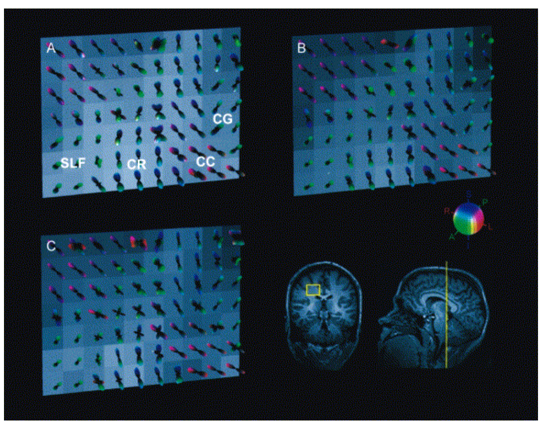
Diffusion tensor:
3 values of D
3 angles

Credit: Larry Frank

Diffusion Imaging Example

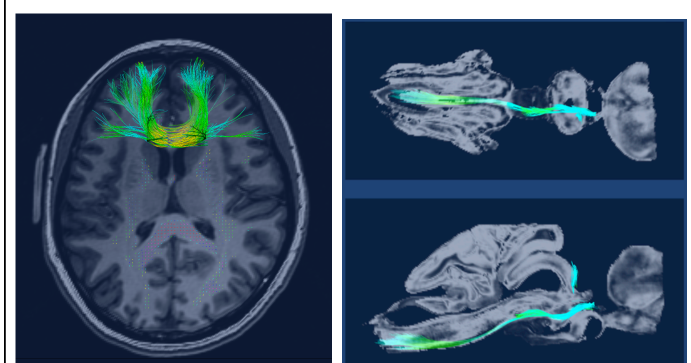


Q-ball imaging



Tuch et al, Neuron 2003

Fiber tract mapping of neural connectivity

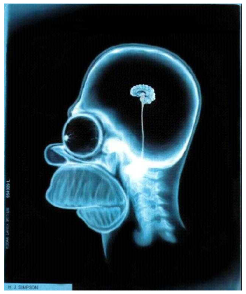


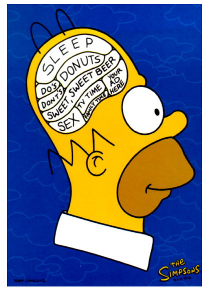
Courtesy of L. Frank

fMRI

MRI studies brain anatomy.

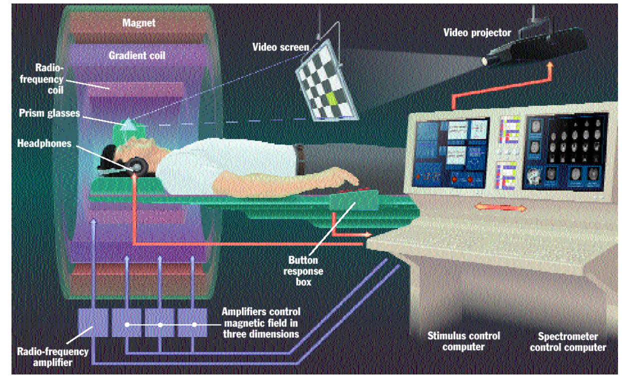
Functional MRI (fMRI) studies brain function.





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

fMRI Setup

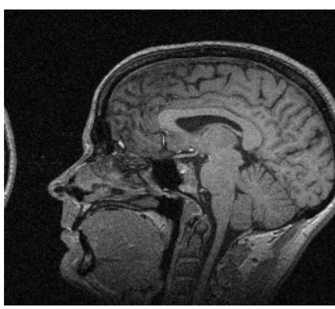


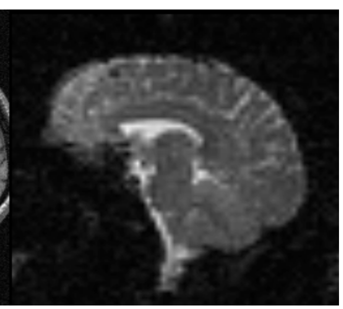
http://defiant.ssc.uwo.ca/Jody_web/fmri4dummies.htm

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

History of Functional MRI

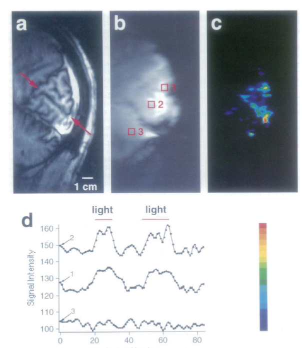


Fig. 1. Intrinsic signal changes in sagittal-brain images produced by photic stimulation. (*a*) Sagittal-slice image of the occipital pole taken with an inversion recovery pulse sequence. The oblique line is oriented along the bank of the calcarine fissure. (*b*) Gradient echo image (FISP sequence; TE = 40 ms) at the same anatomical location. (*c*) Pseudocolor map of the difference in signal intensity between the average of eight images acquired during photic stimulation and eight images taken in the dark. (*d*) Time course of signal-intensity changes (in arbitrary units) for regions indicated by the three boxes outlined in *b*.

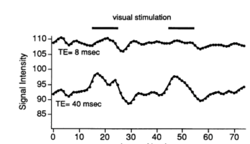
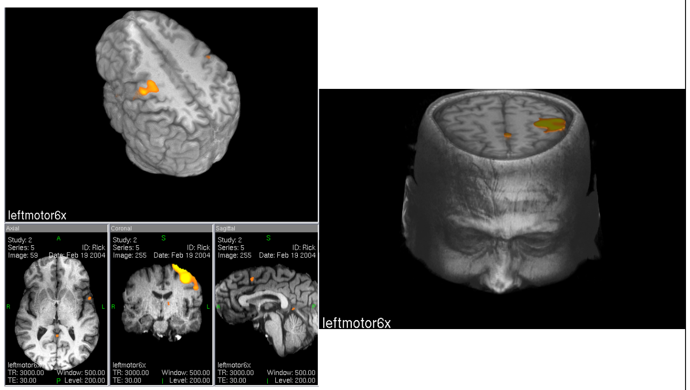


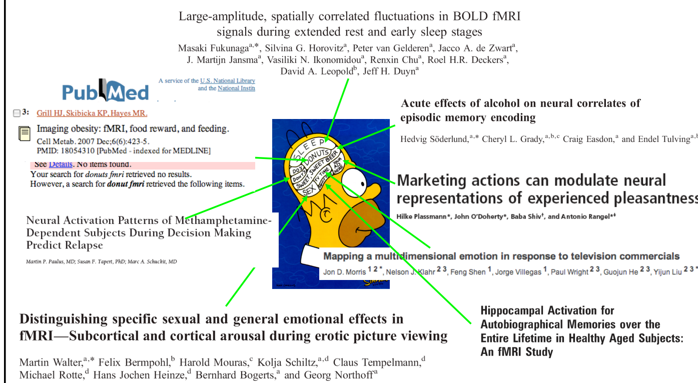
Fig. 3. Reducing TE reduces amplitude of the visual stimulation-induced intrinsic signal change. The time course of intrinsic signal changes observed at a fixed caudal position in primary visual cortex are shown for TE = 40 ms and TE = 8 ms. Other experimental conditions were as in Fig. 2, except that patterned-flash visual stimulation was provided between images 15–25 and 35–55.

Source: Ogawa et al., 1992

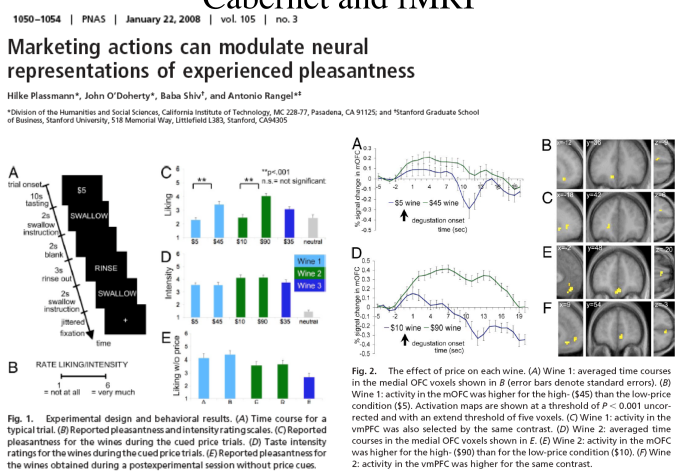
Finger Tapping Task



Functional MRI



Cabernet and fMRI



1050–1054 | PNAS | January 22, 2008 | vol. 105 | no. 3

Marketing actions can modulate neural representations of experienced pleasantness

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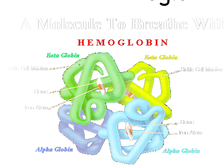
Fig. 1. Experimental design and behavioral results. (*A*) Time course for a typical trial. (*B*) Reported pleasantness and intensity rating scales. (*C*) Reported pleasantness for the wines during the cued price trials. (*D*) Taste intensity ratings for the wines during the cued price trials. (*E*) Reported pleasantness for the wines obtained during a postexperimental session without price cues.

Fig. 2. The effect of price on each wine. (*A*) Wine 1: averaged time courses in the medial OFC voxels shown in *B* (error bars denote standard errors). (*B*) Wine 1: activity in the mOFC was higher for the high- ($45) than the low-price condition ($5). Activation maps are shown at a threshold of $P < 0.001$ uncorrected and with an extend threshold of five voxels. (*C*) Wine 1: activity in the vmPFC was also selected by the same contrast. (*D*) Wine 2: averaged time courses in the medial OFC voxels shown in *E*. (*E*) Wine 2: activity in the mOFC was higher for the high- ($90) than for the low-price condition ($10). (*F*) Wine 2: activity in the vmPFC was higher for the same contrast.

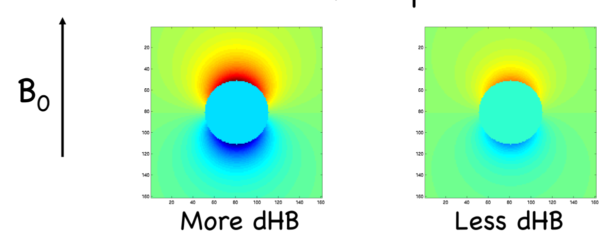
Hemoglobin and Field Inhomogeneities



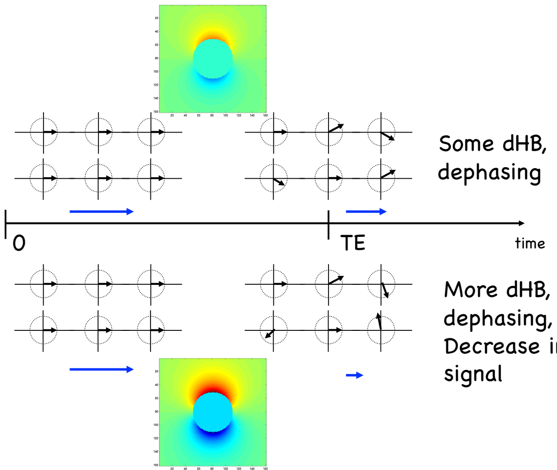
Oxygen binds to the iron atoms to form oxyhemoglobin HbO_2

Release of O_2 to tissue results in deoxyhemoglobin $dHBO_2$

Field Maps



Signal Decay



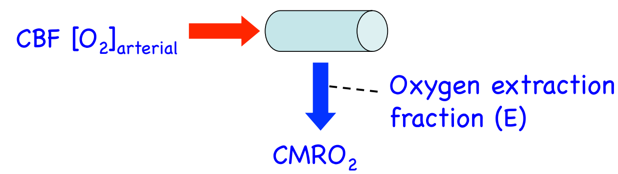
Some dHB, Some dephasing

More dHB, More dephasing, Decrease in MR signal

Blood Flow and Oxygen Metabolism

Cerebral Blood Flow (CBF) measures delivery of blood to brain tissue (units of ml/(g-min))

Cerebral Metabolic Rate of ($CMRO_2$) is the rate of oxygen consumption (units of μmol/(g-min))

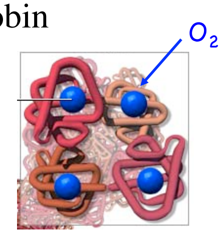


$$CMRO_2 = E \ CBF \ [O_2]_{arterial}$$

Deoxyhemoglobin

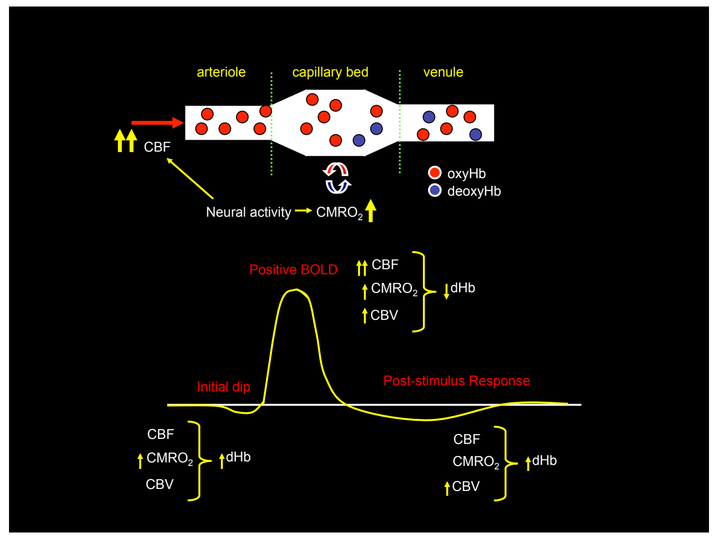
$$[dHB]_{venous} \approx E \ [O_2]_{arterial} \ / \ 4$$

$$= CMRO_2 \ / \ 4CBF$$

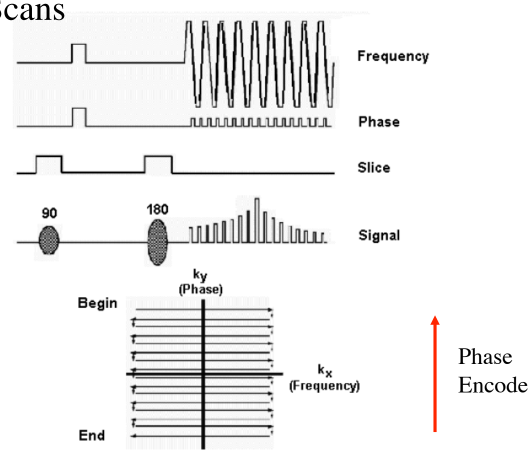


↑ $CMRO_2$ ⟶ ↑ $[dHB]_{venous}$

↑ CBF ⟶ ↓ $[dHB]_{venous}$

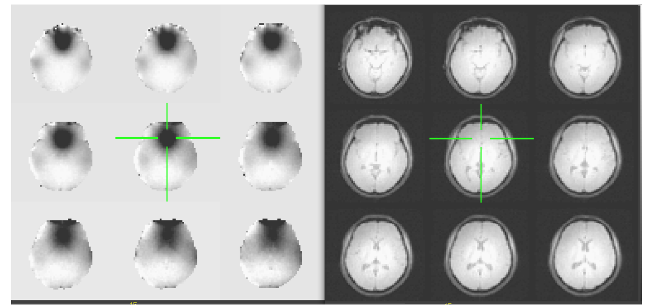


EPI Scans

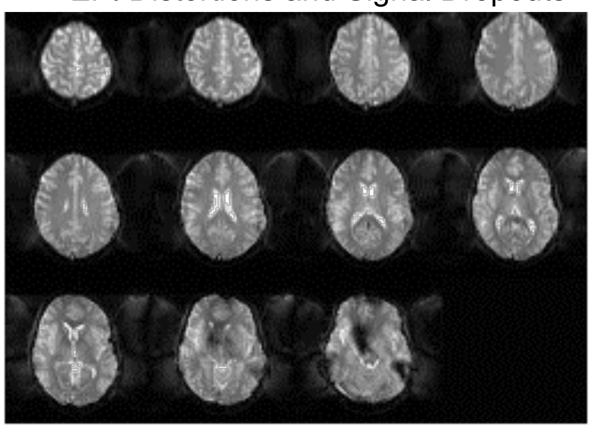


GE Medical Systems 2003

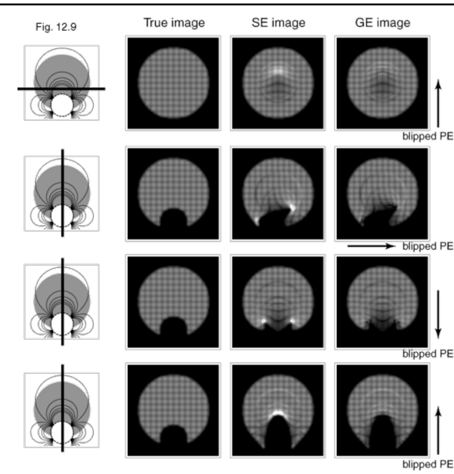
Field Inhomogeneities



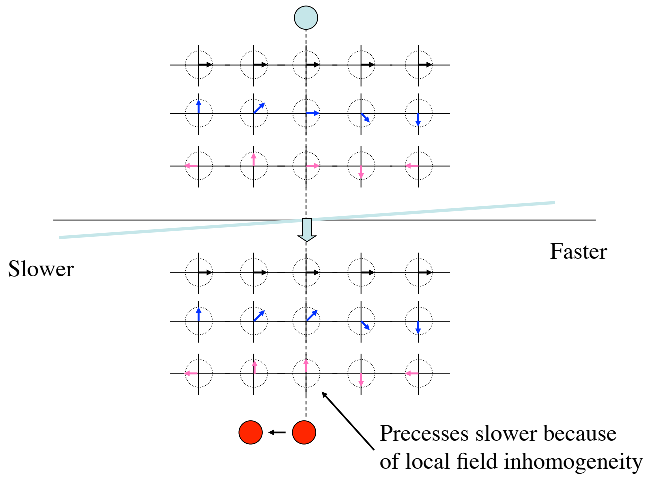
EPI Distortions and Signal Dropouts

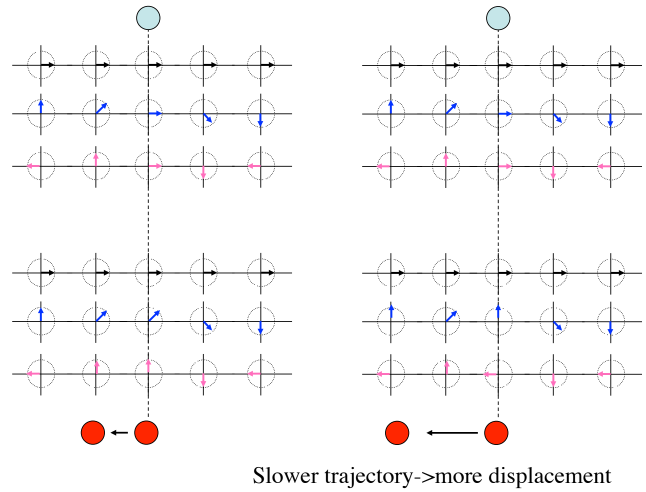


Credit: R. Buxton

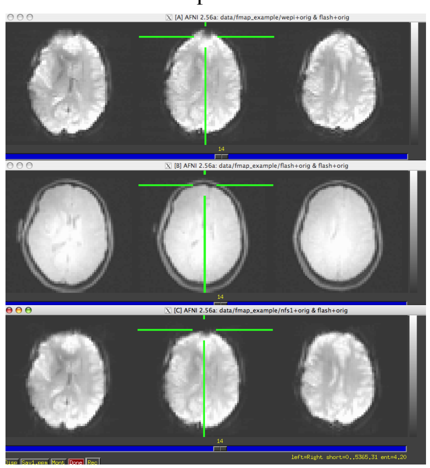


Credit: R. Buxton

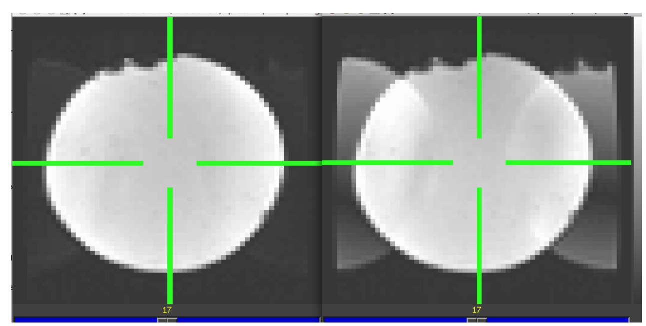




Field Map Correction



Nyquist Ghosts

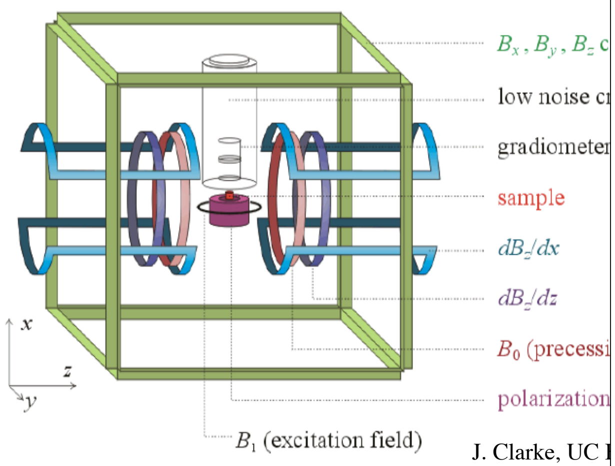


Timeline

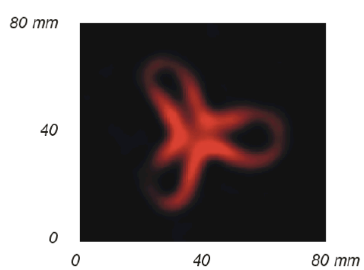
Michael Crichton, 1999

“Most people”, Gordon said, “don’t realize that the ordinary hospital MRI works by changing the quantum state of atoms in your body ... But the ordinary MRI does this with a very powerful magnetic field - say 1.5 tesla, about twenty-five thousand times as strong as the earth’s magnetic field. We don’t need that. We use Superconducting QUantum Interference Devices, or SQUIDs, that are so sensitive they can measure resonance just from the earth’s magnetic field. We don’t have any magnets in there”.

J. Clarke, UC Berkeley

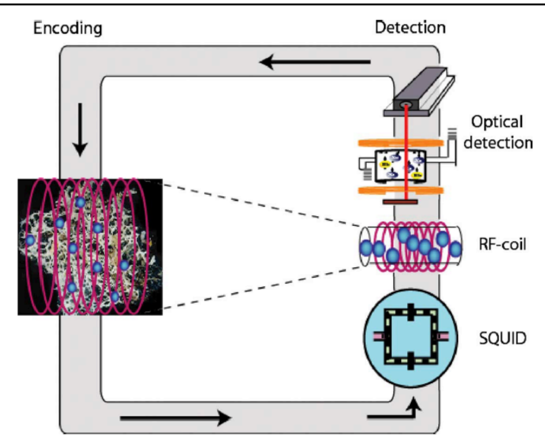


J. Clarke, UC

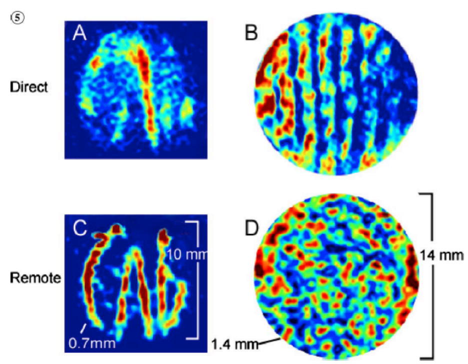


B0 = 132μT
G = 100μT/m
48 projections
1.5 min. acquisition

J. Clarke, UC Berkeley

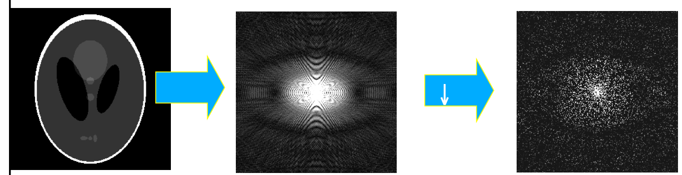


Seeley et al, JMR 2004



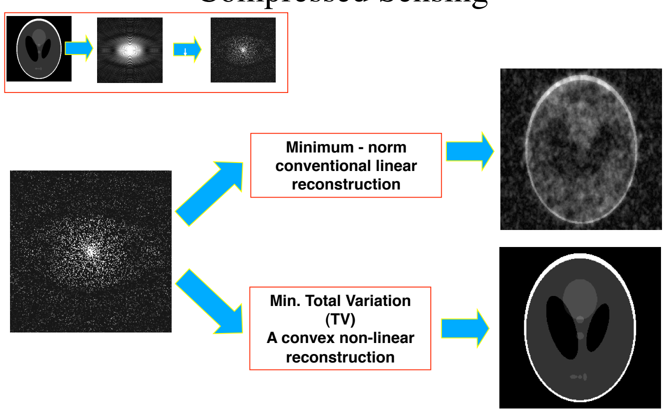
Seeley et al, JMR 2004

Compressed Sensing



Slide Credit: http://www.stanford.edu/~mlustig/

Compressed Sensing



Slide Credit: http://www.stanford.edu/~mlustig/