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 a Moving Spin

\[ q(t) = -\int_0^t \Delta \omega(\tau) d\tau \]
\[ = -\int_0^t \gamma \Delta B(\tau) d\tau \]
\[ = -\int_0^t \gamma \hat{\mathbf{B}}(\tau) \cdot \hat{\mathbf{r}}(\tau) d\tau \]
\[ = -\gamma \int_0^t \left[ G_x(\tau) \dot{x}(\tau) + G_y(\tau) \dot{y}(\tau) + G_z(\tau) \dot{z}(\tau) \right] d\tau \]
Consider motion along the x-axis
\[ x(t) = x_0 + vt + \frac{1}{2}at^2 \]

\[ q(t) = -\gamma \int_0^t G_s(\tau)x(\tau)d\tau \]
\[ = -\gamma \int_0^t G_s(\tau) \left[ x_0 + vt + \frac{1}{2}at^2 \right] d\tau \]
\[ = -\gamma \left[ x_0 \int_0^t G_s(\tau)d\tau + v \int_0^t G_s(\tau)d\tau + \frac{a}{2} \int_0^t G_s(\tau)t^2d\tau \right] \]
\[ = -\gamma \left[ x_0M_0 + vM_1 + \frac{a}{2}M_2 \right] \]

Flow Moment Example

\[ M_0 = \int_0^T G_s(\tau)d\tau = 0 \]
\[ M_1 = \int_0^T G_s(\tau)t d\tau \]
\[ = -\int_0^T G_s d\tau T + \int_0^T \int_0^T G_s d\tau d\tau \]
\[ = G_0 \left[ \frac{T^2}{2} + \frac{T^2}{2} \right] \]
\[ = G_0 \left[ \frac{T^2}{2} \right] = G_0T^2 \]

Phase Contrast Angiography (PCA)

\[ q_1 = -\gamma v_s M_1 = \gamma v_s G_0 T^2 \]
\[ q_2 = -\gamma v_s M_1 = -\gamma v_s G_0 T^2 \]
\[ \Delta q = q_1 - q_2 = 2\gamma v_s G_0 T^2 \]
\[ v_s = \frac{\Delta q}{2G_0 T^2} \]
**PCA example**

- $G_0$

**Aliasing in PCA**

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

$$VENC = \frac{\pi}{\gamma G_0 T}$$

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 $\pi$ radians.

$$\varphi_1 = \frac{v}{VENC_1}$$

$$\varphi_2 = \frac{v}{VENC_2}$$

$$\varphi_1 - \varphi_2 = 2\pi \left( \frac{1}{VENC_1} - \frac{1}{VENC_2} \right)$$

**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

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

Prior to imaging

Relaxed spins flowing in

Saturated spins

Time of Flight Angiography

At TE both the first and second order moments are zero, so both stationary and moving spins have zero net phase.
Cerebral Blood Flow (CBF)

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

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

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

Arterial spin labeling (ASL)

1: Tag by Magnetic Inversion
2: Control - Tag = \( \Delta M \propto \text{CBF} \)
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

PASL / VSASL

\[ \text{Tag} \quad \text{Acquire} \quad \text{Control} \quad \text{Acquire} \]

\[ \text{TI} = \text{Inversion Time} \]

CASL

\[ \text{Tag} \quad \text{Acquire} \quad \text{Control} \quad \text{Acquire} \]

Labeling Time

Post Labeling Delay

Multislice CASL and PICORE

CASL

PICORE

QUIPSS II

Perfusion Images

ASL Time Series

Wait

Image 1

Image 2

Image 3

Image 4

Perfusion Images
Diffusion

\[ \langle \Delta x^2 \rangle = N d^2 = 2D T \]

\( D = \text{diffusivity} \)

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

Diffusing Spins

\[ \Delta B_z(x) \]
\[ \Delta B_x(x) \]

\( D = \text{diffusivity} \)

Credit: Larry Frank

Diffusion Weighting

\[ S \propto e^{-\gamma G_0 \delta^2 DT} = e^{-bd} \]
where \( b = \gamma^2 G_0^2 \delta^3 (T - \delta/3) \)

Signal

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.
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

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**fMRI Setup**

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**fMRI Acquisition**

- High spatial resolution
- High temporal resolution

- **MP-RAGE**
  - Voxel volume: 1 mm$^3$
  - Imaging time: 6 min
  - GE Medical Systems 2003

- **EPI**
  - Voxel volume: 45 mm$^3$
  - Imaging time: 60 msec
  - GE Medical Systems 2003

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**EPI Scans**

EPI (Echo Planar Imaging) scans are used in fMRI to capture rapid changes in brain activity.

- **Frequency**
- **Phase**
- **Slice**
- **Signal**

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Field inhomogeneities also cause the spins to dephase with time and thus for the signal to decrease more rapidly. To first order this can be modeled as an additional decay term.

Precesses slower because of local field inhomogeneity.
Slower trajectory $\rightarrow$ more displacement

Field Map Correction

Nyquist Ghosts

Visual Activation

Source: Kwong et al., 1992

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

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Hemoglobin

Oxygen binds to the iron atoms to form oxyhemoglobin $\text{HbO}_2$

Release of $\text{O}_2$ to tissue results in deoxyhemoglobin $\text{dHBO}_2$

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

Effect of $\text{dHBO}_2$

d$\text{HBO}_2$ is paramagnetic due to the iron atoms. As it becomes oxygenated, it becomes less paramagnetic.

d$\text{HBO}_2$ 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 $\text{dHBO}_2$ 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 $\text{dHBO}_2$ decreases, $T_2^*$ increases and we expect the signal amplitude to go up.
**BOLD Effect**

Blood Oxygen Level Dependent signal

↑ neural activity $\rightarrow$ ↑ blood flow $\rightarrow$ ↑ oxyhemoglobin $\rightarrow$ ↑ $T2^*$ $\rightarrow$ ↑ MR signal

**Basal state**

- normal flow
- basal level [Hb]
- basal $T2^*$
- normal MRI signal

**Activated state**

- increased flow
- decreased [Hb] (lower field gradient around nucleus)
- increased $T2^*$
- increased MRI signal (from lower field gradient)

Source: fMRI and Introduction to BOLD

Thomas Liu, BE280A, UCSD, Fall 2007

http://defiant.science.uwo.ca/lsj_web/fmri4dummies.htm

**Effect of Caffeine on Functional Connectivity**

Thomas Liu, BE280A, UCSD, Fall 2007
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

Seeley et al, JMR 2004
Seeley et al, JMR 2004