BME 50500: Image and Signal Processing in Biomedicine

Lecture 8: Medical Imaging Modalities

MRI

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Filtering
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# Medical Imaging

<table>
<thead>
<tr>
<th>Imaging Modality</th>
<th>Year</th>
<th>Inventor</th>
<th>Wavelength Energy</th>
<th>Physical principle</th>
</tr>
</thead>
<tbody>
<tr>
<td>X-Ray</td>
<td>1895</td>
<td>Röntgen (Nobel 191)</td>
<td>3-100 keV</td>
<td>Measures variable tissue absorption of X-Rays</td>
</tr>
<tr>
<td>Positron Emission Tomography (PET)</td>
<td>1953</td>
<td>Brownell, Sweet</td>
<td>150 keV</td>
<td>SPECT with improved SNR due to increased number of useful events.</td>
</tr>
<tr>
<td>Computed Axial Tomography (CAT or CT)</td>
<td>1972</td>
<td>Hounsfield, Cormack</td>
<td>keV</td>
<td>Multiple axial X-Ray views to obtain 3D volume of absorption.</td>
</tr>
<tr>
<td>Magnetic Resonance Imaging (MRI)</td>
<td>1973</td>
<td>Lauterbur, Mansfield</td>
<td>GHz</td>
<td>Space and tissue dependent resonance frequency of kern spin in variable magnetic field.</td>
</tr>
<tr>
<td>Ultrasound</td>
<td>1940-1955</td>
<td>many</td>
<td>MHz</td>
<td>Measures echo of sound at tissue boundaries.</td>
</tr>
</tbody>
</table>
MRI - Equipment

Source: Joe Gati, photos

Adapted from Jody Culham, http://defiant.ssc.uwo.ca/Jody_web/fmri4dummies.htm
MRI – Basic Recipe

1) Put subject in big magnetic field
   When protons are placed in a constant magnetic field, they precess at a frequency proportional to the strength of the magnetic field (at typical radio frequencies). They also align somewhat to generate a bulk magnetization.

2) Transmit radio waves into subject [about 3 ms]
   Exposure to radio frequency magnetic field will synchronize this precession.

3) Turn off radio wave transmitter
   The coherent precession continues but decays slowly due to interactions with magnetic moments of surrounding atoms and molecules (tissue dependent!)

4) Receive radio waves re-transmitted by subject [10-110ms]
   The coherent precession (oscillation) generates a current in an inductive coil. The detected signal is called magnetic nuclear resonance.

5) Store measured radio wave data vs. time
   Now go back to 2) to get some more data with different magnetic fields and radio frequencies. (here lies the Art of MRI!)

6) Process raw data to reconstruct images

Source: Robert Cox's web site
MRI – Big Magnet

Very strong

1 Tesla (T) = 10,000 Gauss

Earth’s magnetic field = 0.5 Gauss

4 Tesla = 4 x 10,000 ÷ 0.5 = 80,000x Earth’s magnetic field

Continuously on

Main field = $B_0$

Adapted from Jody Culham, http://defiant.ssc.uwo.ca/Jody_web/fmri4dummies.htm
MRI – Nuclear Spin

Nucleus has a quantum mechanical property called “spin” quantized by $I$. ($I=1/2$ for a proton in $\text{H}_2\text{O}$). Spin can be thought of as a spinning mass with an angular momentum $J$.

$$|J| = \frac{\hbar}{2\pi} \sqrt{I^2 + I}$$

Since the particle is electrically charged this spinning will generate a magnetic moment $\mu$:

$$\mu = \gamma J$$

The gyromagnetic ratio $\gamma$ is specific to each nucleus.

As we will see the magnetic fields and radio frequency (RF) are tuned to a specific value of $\gamma$, i.e. to a specific nucleus.
MRI – Nuclear Spin in Magnetic Field

When a spin is placed in a homogeneous external magnetic field $B_0$ it precesses at a frequency $\omega_0$.

$$\omega_0 = \gamma B_0$$

Quantum mechanics however dictates that the valued for the $z$-orientation of $J$ (and $\mu$) can only be:

$$\mu_z = \gamma J_z = \frac{\gamma h}{2\pi} m_I$$

with $m = \pm \frac{1}{2}$ for $I = \frac{1}{2}$.

The effect is analogous to a spinning mass in a gravitational field:
MRI – Nuclear Spin

Properties on nuclei found at high abundance in the body:

<table>
<thead>
<tr>
<th>Nucleus</th>
<th>Atomic Number</th>
<th>Atomic Mass</th>
<th>$\gamma/2\pi$ (MHz/T)</th>
<th>MRI Signal</th>
</tr>
</thead>
<tbody>
<tr>
<td>Proton, $^1$H</td>
<td>1</td>
<td>1</td>
<td>$\frac{1}{2}$</td>
<td>yes</td>
</tr>
<tr>
<td>Phosphorus, $^{31}$P</td>
<td>15</td>
<td>31</td>
<td>$\frac{1}{2}$</td>
<td>yes</td>
</tr>
<tr>
<td>Carbon, $^{12}$C</td>
<td>6</td>
<td>12</td>
<td>0</td>
<td>no</td>
</tr>
<tr>
<td>Oxygen, $^{16}$O</td>
<td>8</td>
<td>16</td>
<td>0</td>
<td>no</td>
</tr>
<tr>
<td>Sodium, $^{23}$Na</td>
<td>11</td>
<td>23</td>
<td>$\frac{3}{2}$</td>
<td>yes</td>
</tr>
</tbody>
</table>

MRI can be performed with odd odd atomic mass (non-zero spin) $^1$H, $^{13}$C, $^{19}$F, $^{23}$Na, $^{31}$P

Most frequent medical imaging is performed with $^1$H (proton) abundant: high concentration in human body high sensitivity: yields large signals

1.5T magnet uses RF at 3.87 MHz for proton imaging.
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Source: Robert Cox's web site
MRI – RF pulse

If we apply in addition to $B_0$ a field component $B_1 (<< B_0)$ in the $x$-direction oscillating at frequency $\omega_0$ the trajectory for $M$ will be:

$$B_x(t) = B_1 \sin(\omega_0 t)$$

This time varying $B_1$ field is applied for a short time (few ms) with an RF coil at the $x$-axis. The final “flip” angle depends on the length of this RF pulse and the strength of $B_1$.

Useful flip angles are:

- $\alpha = 90^\circ$ $M_z$ is converted into $M_y$
- $\alpha = 180^\circ$ $M_z$ is converted into $-M_z$
MRI – RF pulse

The Swing Analogy:

Oscillating spins generate bulk magnetization $M_z$ lined up with $B_0$:

*A bunch of kids are swinging at different swings, all with the same frequency but out of phase. The average weight of the kids is straight down from the pole – it is “aligned” with external gravity.*

RF pulse (oscillating $B_1$) generates transverse $M_x, M_y$ oscillation:

*If parents push a little bit on every swing, in synchrony, and at the natural frequency of the swings, soon all kids are swinging together in phase. The average weight of the kids is now oscillating back and forth, i.e. there is now a oscillating transverse component.*

*How well they are lined up at the end depends on how often and how strong they were pushed. Note that if the parents pushed at a frequency other than the natural frequency of the swings their effort would not amount to much.*
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Source: Robert Cox's web site
MRI – Free Precession - $T_2$ decay

After the RF pulse the system is left only with $B_0$. Any contribution in the transverse direction will precess around $B_0$ at $\omega_0$. Lets now consider the second term:

$$\frac{d M}{dt} = M \times B - \frac{1}{T_2} \begin{bmatrix} M_x \\ M_y \\ 0 \end{bmatrix} - \frac{1}{T_1} \begin{bmatrix} 0 \\ 0 \\ M_z - M_0 \end{bmatrix}$$

This term indicates that $M_x, M_y$ will decay exponentially with a time constant $T_2$. Together with the precession this gives a damped oscillation, e.g. after a $90^\circ$ pulse:

$$\begin{bmatrix} M_x \\ M_y \end{bmatrix}(t) = M_0 e^{-\frac{t}{T_2}} \begin{bmatrix} \sin(-\omega_0 t) \\ \cos(-\omega_0 t) \end{bmatrix}$$
MRI – Free Precession - $T_2$ decay

The reason for this decay process is that each spins each see a slightly different local field around them. Each then oscillates at a slightly different frequency. The spins will be therefore quickly out of step, and the bulk transverse magnetization will disappear.

The local magnetic fields are not the same because:

1. Each spin sees the magnetic field generated by other spins in the molecule. Quantified with $T_2$. (“spin-spin relaxation”)

2. The field $B_0$ is not perfectly homogeneous. Quantified with $T_2^+$ and about 100 shorter than $T_2$.

Total effect is $T_2^*$:

$$\frac{1}{T_2^*} = \frac{1}{T_2} + \frac{1}{T_2^+}$$

$T_2^*$ dominated by $T_2^+$ and is just a few ms.
**MRI – Free Precession - $T_1$ relaxation**

The third term in the Block equation describes the relaxation of the longitudinal magnetization $M_z$:

$$\frac{d M}{d t} = M \times B - \frac{1}{T_2} \left[ \begin{array}{c} M_x \\ M_y \\ 0 \end{array} \right] - \frac{1}{T_1} \left[ \begin{array}{ccc} 0 & 0 & M_z - M_0 \end{array} \right]$$

This is a exponential relaxation back to the equilibrium value $M_0$, e.g. after a $90^\circ$ pulse and a $180^\circ$ respectively:

$$M_z(t) = M_0 \left(1 - e^{-\frac{t}{T_1}}\right)$$

$$M_z(t) = M_0 \left(2 - e^{-\frac{t}{T_1}}\right)$$

This exponential recovery represents the return of the system to its equilibrium condition $M_z=M_0$, whereby the spins loosing energy to the surrounding lattice (“spin-lattice relaxation”){}
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Source: Robert Cox's web site
MRI – RF pulse

Now a oscillating B1 field perpendicular to B0 will be applied at resonant (precession) frequency $\omega_0$
The overall free precession of the bulk magnetization $M$ after RF pulse of $\alpha=90^\circ$ is then

Free precession after $\alpha = 90^\circ$ RF pulse

receiving RF coil
MRI – Free Induction Decay

This precessing magnetization can be measured inductively with an receiver coil tuned to the resonant frequency ($\omega_0=3.87$ MHz for $^1$H). The detected signal is called the Free Induction Decay (FID). If we detect it in with a coil in $x$ and $y$ axis we can construct a complex variable

$$s(t) = s_x(t) + is_y(t) \propto M_x(t) + iM_y(t) = M_{xy}(0)e^{-t/T_2^*}e^{-i\omega_0 t}$$

$M_{xy}(0)$ denotes here the magnitude of the $M_x$, $M_y$ at the end of the RF pulse, i.e. at $t=0$ of the free precession. Its value is dependent of the specific pulse sequence and is affected typically by the decay times $T_1$ and $T_2$.

By modifying the RF pulses and measuring the magnitude of $s(t)$ one can make estimate the decay times $T_1$ and $T_2$. 
T2 – Echo pulse sequence
90° – τ – 180°: the detected signal magnitude is

\[ \propto \exp\left(-\frac{\tau}{T_2}\right) \]

Assignment 8: Generate graphics representing the pulse sequence and FID for inversion recovery and echo pulse.
MRI – Nuclear Magnetic Resonance (NMR)

The decay constants $T_1$ and $T_2$ depend on physical properties of the resonating sample. By measuring the decay constants one can therefore deduce what is in the sample.

In the 70's it was realized that this may be used for medical applications (Damadian)

<table>
<thead>
<tr>
<th>Tissue</th>
<th>T1 (ms)</th>
<th>T2 (ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fat</td>
<td>260</td>
<td>80</td>
</tr>
<tr>
<td>Muscle</td>
<td>870</td>
<td>45</td>
</tr>
<tr>
<td>Brain (gray matter)</td>
<td>900</td>
<td>100</td>
</tr>
<tr>
<td>Brain (white matter)</td>
<td>780</td>
<td>90</td>
</tr>
<tr>
<td>Liver</td>
<td>500</td>
<td>40</td>
</tr>
<tr>
<td>Cerebrospinal fluid</td>
<td>2400</td>
<td>160</td>
</tr>
</tbody>
</table>
## MRI – T1 and T2 images

<table>
<thead>
<tr>
<th>Tissue</th>
<th>MR-T1</th>
<th>MR-T2</th>
<th>CT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bone</td>
<td>dark</td>
<td>dark</td>
<td>bright</td>
</tr>
<tr>
<td>Air</td>
<td>dark</td>
<td>dark</td>
<td>dark</td>
</tr>
<tr>
<td>Fat</td>
<td>bright</td>
<td>bright</td>
<td>dark</td>
</tr>
<tr>
<td>Water</td>
<td>dark</td>
<td>bright</td>
<td>dark</td>
</tr>
<tr>
<td>Brain</td>
<td>anatomic</td>
<td>intermediate</td>
<td>intermediate</td>
</tr>
</tbody>
</table>

Adapted from [http://www.ecf.utoronto.ca/apsc/courses/bme595f/notes/](http://www.ecf.utoronto.ca/apsc/courses/bme595f/notes/)
MRI – How to convert NMR into imaging?

But wait!

How can one generate images?

So far NMR only gives information on the entire sample which is resonating at one frequency $\omega_0$ within the $B_0$ field.

Answer (Lauterbur 1973): Change the $B_0$ field with space and the resonance frequency will change with space. Resonating signal contains multiple frequency components each giving information about a different portion of space!
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6) Process raw data to reconstruct images

Source: Robert Cox's web site
MRI – How to generate images using NMR

Nuclear spins resonate at a frequency proportional to the external magnetic field

$$\omega = \gamma B_0$$

**Basic idea of MRI:** Change the $B_0$ field with space and the resonance frequency will change with space.

$$\omega (r) = \gamma B_0 (r)$$

The detected resonance signal (FID) contains multiple frequency components each giving information about a different portion of space!
MRI – Signal detected in MRI

Recall that the signal due to the bulk magnetization precessing at \( \omega \) detected in the \( x \) and \( y \) coils can be written as:

\[
s(t) = s_x(t) + i s_y(t) \propto M_{xy}(0) e^{-t/T_2^*} e^{-i \omega t}
\]

Signal intensity scales with \( M_{xy}(0) \) - the magnitude of the transverse magnetization at the end of the RF pulse. \( M_{xy}(0) \) is proportional to the number of resonating spins in the material, or the proton density \( \rho(r) \). It is dependent on the tissue and therefore dependent on space \( r \).

**MRI generates images of \( \rho(r) \)!**

\( M_{xy}(0) \) also depends on the specifics of the pulse sequence. By manipulating the pulse sequence MRI can generate images of \( \rho(r) \) that are modulated by physical properties that affect \( T_1 \) or \( T_2 \).
MRI – Signal detected in MRI

The main idea is to apply a $B_0$ field with a magnitude that also depends on space, so that the frequency of the resonance signal relates to space, $\omega(r) = \gamma B_0(r)$:

$$s(t) \propto e^{-t/T_2^*} \rho(r) e^{-i \gamma B_0(r) t}$$

(where we have ignored the effect of $T_1$ and $T_2$). The signal emitted by the entire body is then the sum over space:

$$s(t) \propto e^{-t/T_2^*} \int_{\text{body}} d\mathbf{r} \rho(\mathbf{r}) e^{-i \gamma B_0(\mathbf{r}) t}$$

Note that $B_0(\mathbf{r})$ is parallel to the $z$-axis, only its magnitude may now depend on the location in space $\mathbf{r}$. 
MRI – Signal detected in MRI

For reconstruction it will be useful to define new signal that is 'demodulated' and without the $T_2^*$ decay:

$$S(t) = s(t) e^{t/T_2^*} e^{i\omega_0 t}$$

Define also $\Delta B_z(r)$ as the difference of $B_0(r)$ over main $B_0$:

$$\Delta B_z(r) = B_0(r) - \omega_0 / \gamma$$

With this the MRI imaging equations becomes

$$S(t) = \int_{\text{body}} d\mathbf{r} \rho(\mathbf{r}) e^{-i\gamma \Delta B(\mathbf{r}) t}$$
MRI – $B_0$ gradient, frequency encoding

Lets assume we need spatial resolution in only one direction. For instance $x$. So we want to recover (ignoring $z$ direction for now):

$$g(x) = \int dy \rho(x, y)$$

To do so, we apply a contribution $B_0$ that changes linearly with $x$. The strengths of these 'x-gradient' is given by the constants $G_x$.

$$\Delta B_z(r) = G_x x$$
MRI – $B_0$ gradient, frequency encoding

The imaging equation is now

$$ S(t) = \int d \mathbf{x} g(x) e^{-i \gamma G_x x t} $$

To put this in a more familiar notation lets define a new variable

$$ k_x = \gamma G_x t \quad \gamma = \gamma / 2 \pi $$

$$ S(k_x) = \int d \mathbf{x} g(x) e^{-i 2 \pi k_x x} $$

Evidently the detected signal $S(k)$ is a Fourier transform of $g(x)$, and we can recover it with the inverse Fourier transform.

$$ g(x) = \int d \mathbf{x} S(k_x) e^{i 2 \pi k_x x} $$

This methods is therefore called \textit{frequency encoding}. Obviously we can also apply a $G_y$ gradient and obtain $g(y)$. 
MRI – Axial Reconstruction

By combining \( x, y \) gradients linearly we can get gradients that at an arbitrary orientation \( \phi \):

\[
\Delta B_z (r) = G_x x + G_y y = G_{\phi} \cdot r
\]

\[
G_{\phi} = \begin{bmatrix} G_x \\ G_y \end{bmatrix} = G_{\phi} \begin{bmatrix} \cos \phi \\ \sin \phi \end{bmatrix} \quad r = \begin{bmatrix} x \\ y \end{bmatrix}
\]

The signal we obtain is then a Fourier transform of \( \rho(r) \) along that direction (the orthogonal directions are summed).

\[
k_{\phi} = \begin{bmatrix} k_x \\ k_y \end{bmatrix} = k \begin{bmatrix} \cos \phi \\ \sin \phi \end{bmatrix}
\]

\[
k = \gamma G_{\phi} t
\]

\[
S(t, \phi) = \int d\mathbf{r} \rho(\mathbf{r}) e^{-i \gamma G_{\phi} \cdot \mathbf{r} t}
\]

\[
k_{\phi} = \gamma G_{\phi} t
\]

\[
S(k, \phi) = \int d\mathbf{r} \rho(\mathbf{r}) e^{-i 2\pi k_{\phi} \cdot \mathbf{r}}
\]
MRI – k-space

Signals taken at multiple angles $\phi$ cover the k-space and allow therefore reconstruction (left).

Is there a pulse sequence that can sample the Fourier space evenly as shown on the right so that we can use direct 2D Fourier inverse?
MRI – Slice selection

So far we considered gradients applied after the RF pulse during free precession. A gradient $G_z$ during the RF pulse will select a transversal slice that satisfies the resonance condition: The RF pulse affects the spin precession coherently only if the frequency matches the $B_z$ field. For the rest $M_{xy} = 0$ after $\alpha$ pulse.

\[ M_{xy}(0) = 0 \]

\[ M_{xy}(0) = M_0 \sin \alpha \]

\[ M_{xy}(0) = 0 \]

Only this slice will generate a signal!
MRI – Slice selection

Note that a “hard” RF pulse contains high frequency components. It is therefore less selective in space as a “soft” pulse (sinusoid modulated by a sync functions - $\sin(\omega_0 t)\text{sinc}(\omega t)$):

![Image of RF pulse and selected slice comparison]

- $\text{RF pulse}$
- $\text{Selected slice}$

- hard
- soft

- time
- z axis
MRI – a pulse sequence example

Example for a full pulse sequence with gradient echo and the corresponding path in k-space:

Adapted from http://www.ecf.utoronto.ca/apsc/courses/bme595f/notes/
Echos – refocussing of signal

Spin echo:
use a 180 degree pulse to “mirror image”
the spins in the transverse plane
when “fast” regions get ahead in phase,
make them go to the back and catch up

- measure T2
- ideally TE = average T2

Gradient echo:
flip the gradient from negative to positive
make “fast” regions become “slow” and
vice-versa

- measure T2*
- ideally TE ~ average T2*

A gradient reversal (shown) or 180
pulse (not shown) at this point will
lead to a recovery of transverse
magnetization

TE = time to wait to measure
refocussed spins

MRI – Summary for Magnetic fields

- **Main Magnet**
  - High, constant, Uniform Field, $B_0$.
- **Gradient Coils**
  - Produce pulsed, linear gradients in this field.
  - $G_x$, $G_y$, & $G_z$
- **RF coils**
  - Transmit: $B_1$ Excites NMR signal (FID).
  - Receive: Senses FID.

Adapted from http://www.ecf.utoronto.ca/apsc/courses/bme595f/notes/
MRI – Contrast properties

- The strength of the NMR signal produced by precessing protons in a tissue depends on
  - T1, T2 of the tissue.
  - The density of protons in the tissue.
  - Motion of the protons (flow or diffusion).
  - The MRI pulse sequence used
- In a T1 “weighted” image the pulse sequence is chosen so that T1 has a larger effect than T2.
- Images can also be made to be T1, T2 proton density or flow/diffusion weighted.

Adapted from http://www.ecf.utoronto.ca/apsc/courses/bme595f/notes/
MRI – Contrast, T1, T2

- MRI Contrast is created since different tissues have different T1 and T2.
- Gray Matter: (ms) T1= 810, T2= 101
- White Matter: (ms) T1= 680, T2= 92
- Bone and air are invisible.
- Fat and marrow are bright.
- CSF and muscle are dark.
- Blood vessels are bright.
- Gray matter is darker than white matter.