Basic of Magnetic Resonance Imaging

Seong-Gi Kim

Paul C. Lauterbur Chair in Imaging Research
Professor of Radiology, Neurobiology and Bioengineering
University of Pittsburgh

www.kimlab.pitt.edu
MRI Overview

- First MRI method: Lauterbur 1973
- First clinical MR scanner: GE 1983
- Important diagnostic imaging tool

- Nobel prize on MRI: Lauterbur & Mansfield 2003
Advantages of MRI

- Non-invasive, no ionizing radiation
- Rich contrast mechanisms ($T_1$, $T_2$, density)
- Imaging at different levels

Anatomical → Functional → Molecular
Hardware for MRI
Magnet

Solenoid Coil for Magnet

Earth Magnetic Field: ~0.5 gauss
Refrigerator magnet: ~50 gauss

Human MRI (65 – 90 cm): 1.5 – 9.4 Tesla
Animal MRI (16 – 40 cm): 4.7 – 16.1 T
High-Resolution NMR (54 - 89 mm): 9.4 – 21 T

1 Tesla = 10,000 gauss
Gradient Coils

Generate linear magnetic fields along x, y, and z axis, which can be controlled by computer.

www.magnet.fsu.edu/.../images/mri-scanner.jpg
Radiofrequency (RF) Coils

RF coils are used for excitation of spins and for detection of MRI signals

Birdcage Coil

Surface Coil
Basic NMR
Magnetic Resonance

- Certain atomic nuclei including $^1$H exhibit nuclear magnetic resonance.
- Nuclear “spins” are like magnetic dipoles.

Brian Hargreaves from Stanford
Polarization

- Spins are normally oriented randomly.
- In an applied magnetic field, the spins align with the applied field in their equilibrium state.
- Excess along $B_0$ results in net magnetization.

No Applied Field

Applied Field
Static Magnetic Field

- Longitudinal: \( z \)
- Transverse: \( x, y \)

\( B_0 \)
Precession

• Spins precess about applied magnetic field, $B_0$, that is along $z$ axis.

• The frequency of this precession is proportional to the applied field:

$$\omega = \gamma B$$

Magnetic field strength

Gyromagnetic ratio

Top view
# Nuclei of Biological Interest

<table>
<thead>
<tr>
<th>Nucleus</th>
<th>Net Spin</th>
<th>$\gamma$ (MHz/T)</th>
<th>Natural Abundance</th>
</tr>
</thead>
<tbody>
<tr>
<td>$^1$H</td>
<td>1/2</td>
<td>42.58</td>
<td>99.99%</td>
</tr>
<tr>
<td>$^{31}$P</td>
<td>1/2</td>
<td>17.25</td>
<td>100%</td>
</tr>
<tr>
<td>$^{23}$Na</td>
<td>3/2</td>
<td>11.27</td>
<td>100%</td>
</tr>
<tr>
<td>$^{13}$C</td>
<td>1/2</td>
<td>10.71</td>
<td>1.11%</td>
</tr>
<tr>
<td>$^{14}$N</td>
<td>1</td>
<td>3.08</td>
<td>99.63%</td>
</tr>
</tbody>
</table>
Perturbation of Magnetization

Perturb magnetization with radiofrequency pulses

Flip angle \( \theta = \gamma B_1 \cdot \text{Pulse Duration} \)
Excitation of Spins

- Spins only respond to RF at a frequency matched to the Larmor or precessional frequency!

- RF pulses \( (B_1) \) are induced by the RF coil aligned orthogonal to \( B_0 \). \( B_1 \ll B_0 \)

- Spins that were previously aligned along \( B_0 \) (or z direction) precess around x-axis, or the direction of the newly applied field, \( B_1 \)
• Precessing spins cause a change in flux ($\Phi$) in a transverse receive coil.

• Flux change induces a voltage across the coil.
Signal Reception

• The detected signal is at the Larmor frequency

• One can only receive the signal when axis of detection coil is perpendicular to $B_0$

• The signal loses by dephasing of spins
Dephasing

- Loss of Mxy is due to spin de-phasing

- After 90° RF pulse is off, spins are all lined up in same direction

- During their precession in x-y plane, they begin to wonder away from each other and their collective contribution into the detector diminishes.
Relaxation

- Magnetization returns exponentially to equilibrium:
  - Longitudinal *recovery* time constant is $T_1$ (spin-lattice relaxation time)
  - Transverse *decay* time constant is $T_2$ (spin-spin relaxation time)
Relaxation

- $T_1$ and $T_2$ are due to independent processes
- Generally $T_2 < T_1$
- Dependent on tissue type and magnetic field
$T_2$ Contrast

$T_2$ value is intrinsic to type of tissue

e.g., at 1.5T
- Gray matter: 100 ms
- White matter: 80 ms
- Cerebral spinal fluid: 2000 ms

http://www.med.harvard.edu/AANLIB/home.html
$T_2$ Contrast

Short Echo-Time

Long Echo-Time

CSF

White/Gray Matter

Signal

Time
$T_1$ value is intrinsic to type of tissue

e.g., at 1.5T

- Gray matter: 900 ms
- White matter: 600 ms
- Cerebral spinal fluid: 4000 ms

Caudate nucleus

Putamen

Thalamus
$T_1$ Contrast

Short Repetition

Long Repetition

Signal vs. Time

White/Gray Matter vs. Time

CSF vs. Time
Larmor frequency is sensitive to local field perturbations - these give rise to frequency shifts, and/or a distribution of frequencies.

\[ \omega = \gamma (B_0 + \Delta B) \]

where:
\[ \Delta B = \text{any field perturbation due to:} \]
- chemical shift (i.e., NMR spectroscopy)
- external magnetic field gradient (i.e., imaging)
\( T_2^* \) (apparent transverse relaxation time)

- The net decay constant of \( M_{xy} \) due to both the spin-spin interaction and external magnetic field inhomogeneities (\( \Delta B \)) is called \( T_2^* \)

- \( \frac{1}{T_2^*} = \frac{1}{T_2} + \gamma \Delta B \)

- \( T_2 \geq T_2^* \)
How to measure $T_2$

- How to acquire the MRI signal without the dephasing contribution from static external magnetic field inhomogeneities ($\Delta B$), or $T_2^*$ effects

- Solution: use a spin-echo!

- Spin-echo signal detection is one of the most common methods used in MRI
Spin Echo (two spins)

$t = 0$ (after $90^\circ$ pulse)

Spin-echo

$180^\circ$ pulse along $x'$
Spin Echo

• The amazing property of the spin-echo is that the dephasing contribution from static magnetic field inhomogeneity is refocused and thus eliminated.

• The amplitude of the spin-echo is independent of $T_2^*$, but depends on $T_2$

• One cannot refocus dephasing due to the microscopic spin-spin interaction ($T_2$)
Magnetic Field Gradients

• Spatial information is obtained by the application of magnetic field gradients (i.e. a magnetic field that changes from point-to-point).

• Gradients are denoted as Gx, Gy, Gz, corresponding to the x, y, or z directions. Any combination of Gx, Gy, Gz can be applied to get a gradient along an arbitrary direction (gradients are vector quantities).

• Depending on the gradient’s function, these gradients are called
  – Slice-select gradient
  – The read or frequency-encoding gradient
  – The phase-encoding gradient
Slice Selection Gradient

- Gradient coils provide a linear variation in $B_z$ with position.
- Result is a resonant frequency variation with position.

$$\omega = \gamma (B_0 + B_z)$$
Selective Excitation

Slice
(position & thickness)

Slope = \frac{1}{\gamma G}

RF Pulse
(Resonance Freq and Bandwidth)

Thickness = \frac{BW}{\gamma B_z}
RF Pulse for Excitation

• The bandwidth of an RF pulse depends on its length and shape.

• Fourier Transform of a RF pulse displays bandwidth.

• A RF pulse with a sinc profile is commonly used in MRI for slice selection.
Frequency Encoding

- After having defined a slice through the subject, we need to resolve features along the other two directions (x and y) using frequency-encoding (along x) and phase encoding (along y).

- A smallest volume element in this slice is called a “voxel”.

- The frequency encoding gradient is applied when we “read-out” signals.
Image Acquisition

- Gradient causes resonant frequency to vary with position.
- Receive sum of signals from each spin.
Image Reconstruction

- Received signal is a sum of “tones.”
- The “tones” of the signal are intensities of objects.
- This also applies to 2D and 3D images.
Readout Example

H$_2$O in tubes

G

time domain

FT

frequency domain
Phase Encoding

- Phase encoding resolves spatial features in the vertical direction (y) by using the phase information of precessing spins.

- To get enough data to make an image, we need to repeat the phase encoding process many times, each time with a different strength of phase encoding to impart a different phase angle to the voxel.
Number of Phase Encoding Step

- The number of phase encoding steps is equal to the number of rows in the image (i.e., the resolution in the y-direction).
- The phase shift between adjacent rows is \( \Delta \theta = \frac{360^\circ}{\text{# rows}} \).
Pulse Sequences

- Excitation and imaging are separate.

- Pulse sequence controls:
  - RF excitation
  - Gradient waveforms
  - Acquisition
  - Reconstruction information as well.
1D-Pulse Sequence

Excitation

Readout

RF

Gz

Gx

Acq.
1D-Pulse Sequence – Detailed!

- RF: Phase, Modulation Frequency
- $G_z$: Finite amplitude, slew rate
- $G_x$
- Acq.: • Demodulation frequency, phase • Sampling rate and duration
2-D Image Sequence

RF

$G_z$

$G_y$

$G_x$

Acq.

Excitation  Phase-encoding  Readout
2D Image Reconstruction

$2D$ Image Reconstruction

$\mathbf{FT}$

$k_y$ (phase-encoding)

$k_x$ (readout)

Frequency-space (k-space)

Image space
Resolution

- Image resolution increases as higher spatial frequencies are acquired.
k-Space Trajectories

2D Fourier Transform

Echo-Planar

Spiral