Introduction to Medical Imaging

MRI Imaging

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What We have Learned So Far…

Steady state:
- with just an axial B-field, the nuclei only spin along the z-direction which cannot be measured
- the nuclei also spin in an unsynchronized fashion, which sets their net effect to zero

Disturbed state:
- a transversal RF pulse (at Larmor frequency $\omega_0$) can bring the nuclei spins out of their z-alignment, creating a transversal component in the net magnetization vector
- this transversal RF pulse also synchronizes the spins
- this component can then be measured using (possibly the same) RF coils, whose axes are in the transversal plane (the xy-plane)

A remaining challenge is:
- we receive a single sinusoidal signal (at Larmor frequency $\omega_0$) characterized by amplitude A and phase shift $\theta_s$ → 2 knowns
- what we want is the density of $nx \cdot ny \cdot nz$ individual voxels → $nx \cdot ny \cdot nz$ unknowns
- how do we solve such an equation?
The General Idea

Before going into the math, we shall pursue a more intuitive route to provide an explanation of the used approach

- see also “The basics of MRI” by Joseph Hornak
  [http://www.cis.rit.edu/htbooks/mri](http://www.cis.rit.edu/htbooks/mri)

Recall that the Larmor frequency at which the nuclei spin is dependent on the magnetic field: \( \omega_0 = \gamma B_0 \)

Then how about this:

- if we could simply change the local frequency at each nuclei (or better, voxel) then we could select a voxel just by its frequency
- we would then measure its amplitude to get its density
- we learnt that frequency space is a good space for this

So how is this done?

- we can influence the local frequency by the local magnet field
- the local magnet field can be altered by adding a varying magnet field to the large existing one, \( B_0 \)
Slice Selection

Step 1: slice coding

• select the slice $s$ by coding the nuclei in that slice with a specific $\omega_s$
• this is done by adding a magnetic field linearly rising along $z$ (called gradient field)
• now only the voxels in the selected slice spin at $\omega_s$

Step 2: RF pulse at $\omega_s$

• this will only bring the nuclei in slice $s$ into the transversal plane and sync them
• we could then measure a signal but it would still be a composite of all nuclei in the slice

Hence, more tagging is needed within the slice

next set of slides based on web site
Basics of MRI, by J. Hornak (RPI)
The Phase Encoding Concept

Once the slice has been selected we can impose another magnetic gradient field, say along $x$

- now the nuclei spin at different frequencies, varying along $x$

After releasing the gradient, the nuclei spin at the slice’s Larmor frequency again, but an $x$-dependent phase shift angle remains

- we need a method to measure this phase shift
- we also still need a method to distinguish different rows $y$
After slice selection all nuclei in the slice spin at the same rate.

Apply a phase encoding (along x)
- after gradient release, the column nuclei are distinguished by phase.

Apply another gradient field (now along y)
- do not release → frequency encoding
- now nuclei within a column can be distinguished by frequency.
Decoding

We could measure the RF signal and perform a Fourier transform

- this would give us $ny$ frequencies
- however, we would like to measure $nx \cdot ny$ voxels

Need to generate more equations

- we can get these by performing $nx$ different phase encodings
- for each phase encoding we use a gradient field with a different slope

![Set of gradient fields along x](image)

- the further out a voxel is on this slope (x-coordinate), the greater the additional spin distance it can traverse, and the greater the frequency of the additional spin

- each is then followed by a frequency encoding

Let’s examine this by ways of a few examples
Voxel Decoding: Example 1

spatial domain: only one voxel is turned on

Phase angles of the Fourier transforms along the frequency encoding direction

Extrapolation into continuous signal (just for illustration)

Fourier transform along the phase encoding direction yields voxel position:

$n$ RF measured responses
Voxel Decoding: Example 2

spatial domain: only one voxel is turned on

Phase angles of the Fourier transforms along the frequency encoding direction

Extrapolation into continuous signal (just for illustration)

Fourier transform along the phase encoding direction yields voxel position:

$n$ RF measured responses
Voxel Decoding: Example 3

spatial domain: only one voxel is turned on

$n$ RF measured responses

Phase angles of the Fourier transforms along the frequency encoding direction

Extrapolation into continuous signal (just for illustration)

Fourier transform along the phase encoding direction yields voxel position:
Voxel Decoding: Example 4

spatial domain: only one voxel is turned on

$n$ RF measured responses

Phase angles of the Fourier transforms along the frequency encoding direction

Extrapolation into continuous signal (just for illustration)

Fourier transform along the phase encoding direction yields voxel position:
Image Generation

The spectra have amplitudes as well and an image can be generated

or more complex:
Gradient Coils

Field gradients, generated by coils, are essential in MRI.

Examples:
- Surface coils
- Volume (quadrature) coil
- Head volume coil

x-direction:

y-direction:

z-direction:
More Formally… Slice Selection

Usually one selects a slice perpendicular to the z-axis
(but any slice orientation is possible)

Use linear magnetic field gradient $G_z$:

$$G = (G_x, G_y, G_z) = (0, 0, \frac{\partial B_z}{\partial z})$$

• the Larmor frequency of a slice is then:

$$\omega(z) = \gamma (B_0 + G_z z)$$

• hence, a slice of thickness $\Delta z$ will resonate at frequencies:

$$\Delta \omega = \gamma G_z \Delta z$$

• and a RF pulse of freq. bandwidth $BW$ resonates a slice of width:

There is a lower bound on $\Delta z$:

• technical and safety limits on gradient strength $G_z$
• a very thin slice has too low SNR
• typically: $\Delta z = 2 \text{ mm} \ (B_0=1.5\text{T}), \ \Delta z = 3 \text{ mm} \ (B_0=1\text{T})$
Dynamic cardiac sequence

- four cardiac chambers together with the heart valves in a plane parallel to the cardiac axis
The slice selection RF pulse puts all nuclei within $\Delta z$ into a similar synchronized transverse spin (resonance)

- now all nuclei within the slice’s $(x,y)$-plane have a transverse component
- these must be further encoded for spatial localization
- all nuclei rotate at $\omega_0$, in a synchronized fashion

Let’s assume, for now, that we only have one $x$-beam of voxels

For encoding, we apply a field gradient $G_x$ along $x$:

- this changes the transverse component frequency to:
  \[ \omega(x) = \gamma G_x x \]

- in complex notation:
  \[ M_x(t) = M_x(0)e^{-i\gamma G_xt} \]
To collect signals from all x-beam voxels, we must integrate across the entire x-beam. The detected signal is:

\[
s(t) = \int_{x=-\infty}^{+\infty} \rho(x) e^{-i\gamma G_x x t} \, dx
\]

\(\rho(x)\) is the (transverse) net magnetization density in \(x\).

Defining \(k_x = \frac{\gamma}{2\pi} G_x t\), we get:

\[
s(t) = S(k_x) = \int_{x=-\infty}^{+\infty} \rho(x) e^{-i2\pi k_x x} \, dx
\]

MRI measures: \(f(x) = \rho(x)\)

Recall 1D Fourier Transform:

\[
F(k_x) = \int_{x=-\infty}^{+\infty} f(x) e^{-i2\pi k_x x} \, dx
\]

We can interpret this as:

\[
s(t) = S(k_x) = F\{\rho(x)\}(k_x)
\]

- \(k_x\) grows as long as the gradient is active (as \(t\) increases)
A generalization of what we have just seen

- position vector \( r = (x, y) \) and magnetization density \( \rho(x, y) \) are now 2D vectors
- the angular frequency is given by the dot product: \( \omega(r) = \gamma (\vec{G} \cdot r) \)
- the measured signal becomes:

\[
s(t) = \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} \rho(x, y) e^{-i\gamma (\vec{G} \cdot r)t} \, dx \, dy
= \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} \rho(x, y) e^{-i\gamma (G_x x + G_y y)t} \, dx \, dy
\]

- recall the 2D Fourier Transform (and \( k_x = \frac{\gamma}{2\pi} G_x t \) \( k_y = \frac{\gamma}{2\pi} G_y t \))

\[
F(k_x, k_y) = \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} f(x, y) e^{-i2\pi (k_x x + k_y y)} \, dx \, dy
\]

- thus, MRI allows us to collect data along any line in \( k\)-space:
The K-Theorem (2)

Generalizing further into 3D:

\[
s(t) = \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} \rho(x,y,z) e^{-i\gamma (\vec{G} \cdot \vec{r}) t} \, dx \, dy \, dz
\]

Taking into account relaxation effects \( g(t,T_1) \) and \( h(t,T_2) \):

\[
s(t) = \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} \rho(x,y,z) g(t,T_1) h(t,T_2) e^{-i\gamma (\vec{G} \cdot \vec{r}) t} \, dx \, dy \, dz
\]

MRI seeks techniques where the returned values at \((x,y,z)\) all have the same relaxation effects

- the relaxation effects allow MRI to obtain images of different contrasts, since different types of tissues have different \( g(t,T_1) \) and \( h(t,T_2) \)
- thus, an MRI image is not (necessarily) just a proton density \((\rho)\) image (but this can be obtained as well)
The K-Theorem (3)

We have seen that the time signal $s(t)$ is equivalent to the Fourier Transform (FT) of the desired image

- as mentioned, this data collection space is called \emph{k-space}
- an inverse FT will produce the desired image

![Diagram showing k-space image transforming to spatial image through 2D Fourier transform.](image)
Dephasing Phenomena

Dephasing occurs when the vectors precess at different Larmor frequencies
• this results in a reduced (and noisy) receiver coil signal

Distinguish between three types of dephasing phenomena:

1. Spin-spin interaction:
   • irreversible, described by time constant $T_2$

2. Magnetic field inhomogeneities:
   • reversible, described by time constant $T_2^* < T_2$
   • due to inhomogeneous magnetic field and differences in magnetic susceptibility of the tissues (for example, iron is very susceptible and can change the local field significantly)
   • can be undone by a 180° RF pulse

3. Magnetic field gradients:
   • reversible, time constant $T_2^{*''} < T_2^*$
   • undone with reversed gradient fields
Rephasing: Runners Equivalent

Rephasing is the process of undoing the effects of dephasing

- two types of rephasing, illustrated by the famous runner example

180° RF pulse

Gradient reversal

running at different speeds

runners transposed

from: Frits Thorsen (U Bergen)
Rephasing: 180 RF Pulse

- **TE** = Time to Echo
- **TR** = Repetition Time (can also be chosen for different effects)

large **TR** ensure that all protons are aligned with the z-axis before the next RF pulse

acquired signal, reduced by $T_2$ effects
degree of reduction depends on choice of **TE**

dephase time = rephase time

dephasing

rephasing
Rephasing: Gradient Reversal

Opposite field gradients of same length and strength
TR and TE can be varied to bring out proton density (PD), T1, and T2 properties of the tissue

Recall the Bloch relaxation formula: \( M_{xy} = M_{xy0} (1 - e^{-\frac{TR}{T1}}) e^{-\frac{TE}{T2}} \)

where \( M_{xy0} \) is the transversal component at \( t=0 \)

MRI seeks to bring out tissue contrast as manifested by differences in PD, T1, or T2 properties (T1/TR, T2/TE ratios)

• choose TE and TR according to the T1 and T2 constants of the target and adjacent tissues
• pick the imaging protocol that brings out these contrasts best
• this is called T1, T2, PD weighting
Examples

T1-weighting typically brings out anatomy better
T-2 weighting shows pathologies better
PD is used when T1/T2 densities are similar
Often, all are obtained and possibly fused

proton density  T1-weighted  T2-weighted  T1/T2 fused
Long TR

- to give all tissue time to recover (back to \( M_{z0} \)) and thus eliminate T1 weighting

Long TE

- to give T2 weighting time to develop

\[
M_{xy}^{T2} = M_{xy0} (1 - e^{-\frac{TR}{T1}}) e^{-\frac{TE}{T2}} \Rightarrow M_{z0} e^{-\frac{TE}{T2}}
\]
T2 Weighting

Flip Magnetization

Long TE Time

Long TR Time

B_0

M_{xy}

Brain

CSF
T2 Weighting: Examples

TR = 3070ms   TE = 92ms

TR = 4000ms
TE = 132ms
**T1 Weighting**

**Short TR**
- to bring out differences in T1 between tissues

**Short TE**
- so differences in T2 do not have time to appear

\[
M_{xy}^{T1} = M_{xy0}(1 - e^{-\frac{TR}{T1}}) e^{-\frac{TE}{T2}} \Rightarrow M_{z0}(1 - e^{-\frac{TR}{T1}}) \Rightarrow M_{z0} \frac{TR}{T1}
\]

protons have not recovered to \(M_{z0}\) yet therefore, the 90° RF pulse will not be able to spin them fully into the xy-plane
T1 Weighting: First Time Around

Flip Magnetization

Short TE Time

Short TR Time

Brain

CSF
T1 Weighting: Second Time Around

Flip Magnetization

Short TE Time

Short TR Time

$B_0$

$M_{xy}$

Brain

CSF
T1 Weighting: All Subsequent Rounds

Flip Magnetization

Short TR Time

Short TE Time

Brain

CSF
T1 Weighting: Examples

TR = 525ms  TE = 15ms

TR = 562ms  TE = 20ms
<table>
<thead>
<tr>
<th>Substance</th>
<th>T1 Weighted</th>
<th>T2 Weighted</th>
</tr>
</thead>
<tbody>
<tr>
<td>Water/Vitrious/CSF</td>
<td>black</td>
<td>light grey or white</td>
</tr>
<tr>
<td>Fat</td>
<td>white</td>
<td>light grey</td>
</tr>
<tr>
<td>Muscle</td>
<td>grey</td>
<td>grey</td>
</tr>
<tr>
<td>Air</td>
<td>black</td>
<td>black</td>
</tr>
<tr>
<td>Fatty bone marrow</td>
<td>white</td>
<td>Light grey</td>
</tr>
<tr>
<td>Brain white matter</td>
<td>light grey</td>
<td>grey</td>
</tr>
<tr>
<td>Brain grey matter</td>
<td>grey</td>
<td>Very light grey</td>
</tr>
</tbody>
</table>

T1 weighted
short TR = 500ms
short TE < 30ms

T2 weighted
long TR = 1500ms (3xT1 max)
long TE > 80ms
T1 vs. T2 Weighting (2)

TR = 562ms
TE = 20ms

TR = 4000ms
TE = 132ms
T1 vs. T2 Weighting (3)

TR=525ms TE=15ms

TR=2500ms TE=85ms
Proton Density Weighting

Long TR
- to give all tissue time to recover and thus eliminate T1 weighting

Short TE
- to give no T2 weighting

\[ M_{xy}^{PD} = M_{xy0} \left(1 - e^{-\frac{TR}{T1}}\right) e^{-\frac{TE}{T2}} \Rightarrow M_{z0} \]
Proton Density vs. T2 (1)

TR = 3070ms
TE = 15ms

TR = 3070ms
TE = 92ms
Proton Density vs. T2 (2)

TR = 3070ms
TE = 15ms

TR = 3070ms
TE = 92ms
Knee with Pigmented Villonodular Synovitis (PVNS)

- example shows that PD and T1 weighting creates contrast for complementary features
Use TR to maximize contrast among tissues of interest

from: Frits Thorsen (U Bergen)
Contrast Dependence on TE

Use TE to maximize contrast among tissues of interest

from: Frits Thorsen (U Bergen)
Other Techniques

Gadolinium injection
- can reduce T1 in certain tissues and therefore enhance T1 results

Inversion recovery
- can provide very strong contrast between tissues with different T1
- involves successive 180° and 90° pulses
- tends to be slower than regular T1

FLAIR=Fluid Attenuated Inversion Recovery
- used often to enhance lesions inside the brain in areas close to the fluid surrounding it

And many more…
Sequences: Spin-Echo

Phase shift: \( \phi(y) = \gamma G_y y T_{\text{phase-encoding}} \)

In k-space: \( G_y = mg_y \) where \( m \in \mathbb{Z} \) and \( g_y = \text{constant} \)

\[ k_y = \frac{\gamma}{2\pi} mg_y T_{\text{phase-encoding}} \]
Spin-Echo: Practical Considerations

**Half-Fourier space acquisition:** possible since spin density is a real function

Although faster, this is not done in practice since a full k-space acquisition improves SNR

**Truncated Fourier space acquisition:** requires fewer phase encoding steps

Also faster, but reduces the spatial resolution in the y-direction
Gradient-Echo Pulse Sequences

Slow acquisition speed is a major drawback of Spin-Echo
- particularly in $\rho$ and $T_2$-weighted protocols where TR is long to minimize $T_1$ effects

Gradient-Echo is an alternative technique designed to overcome these limitations:
- flip angle is less than 90° (usually 20° - 60°)
- no spin-echo due to absence of 180° rephasing pulse
- rephasing only by gradient reversal $\rightarrow$ signal characteristics due to $T_2^*$, not $T_2$

Could obtain images using the same TE, TR values than Spin-Echo
- but this would be not faster than Spin-Echo
- the signal would also be lower due to $T_2^*$ effects
- instead perform faster earlier acquisition $\rightarrow$ measure $T_1$ effects

![Diagram showing signal intensity and TR relationship with small flip angles, short TRs returning a larger signal](image)
Comparison: 180° RF Pulse vs. Gradient Echo

180° RF compensates for magnetic field inhomogeneity effects
- example: ferromagnetic particles in a knee joint

Spin-Echo compensates for dephasing
- surrounding areas OK (with some geometric distortion)

Gradient-Echo cannot compensate for dephasing:
- surrounding areas have no measurable signal
Gradient-Echo: FLASH

rephase $T_2^*$ effects

“spoiler”: dephases all remaining transverse components

no k-space flip since no $180^\circ$ RF pulse

slice selection

phase-encoding

frequency-encoding

FLASH = Fast Low Angle Shot

typical image size (like in Spin-Echo): $256^2$

16 short axis cardiac images acquired with 2D FLASH Gradient-Echo sequence
Three-Dimensional Imaging

Direct 3D imaging overcomes the problem with thin-slice selection that plagues 2D methods

Needed for many radiological examinations
  • wrist, knee, ankle, …

Requires a second phase-encoding gradient

\[ \phi(y, z) = \gamma (mg_y yT_{\text{phase-encoding}} + ng_z zT_{\text{slab-selection}}) \]

The setting of \( n \) selects the slice

Reconstruction by ways of a 3D FT
  • obtain 16, 32, 100, … 256\(^2\) pixel slices
  • assuming a slab thickness of 32 mm, can get 32 1 mm thin slices (impossible with 2D imaging)
  • better SNR since each excitation selects all spins within the thick slice and not just a thin slice
  • slower speed due to the multiple slices
Acquisition and Reconstruction Time

Clinical practice:

• high-quality images useless if it takes tens of minutes to acquire them
• reconstruction time fast since FT implemented in special chips

Acquisition time TA determined by:

• total number of excitations
• multiplied by the interval between two successive excitations TR

More concretely:

• 2D pulse sequences: $T_{A_{2D}} = N_{ph} N_{ac} TR$
• 3D pulse sequences: $T_{A_{3D}} = N_{ph} N_{sl} N_{ac} TR$

• $N_{ph}$: number of phase encoding steps
• $N_{ac}$: number of acquisitions (to improve SNR sometimes multiple acquisitions are taken)
• $N_{sl}$: number of slices (for 3D)
Assume a 32 x 256 x 256 volume:

- $N_{ac} = 1$, $N_{sl} = 32$, $N_{ph} = 256$
- with Spin-Echo, $TR = 2000$ ms $\rightarrow T_{A3D} = 4$ hours (infeasible)
- with Gradient-Echo, $TR = 40$ ms $\rightarrow T_{A3D} = 6$ minutes (acceptable)
Fast Imaging Sequences

Recall acquisition time equation:

\[ TA_{2D} = \frac{N_{ac} N_{ph} TR}{ETL} \]

Could get faster by

- decreasing TR (GE (gradient echo) vs. SE (spin echo) sequences)
- decreasing \( N_{ac} \) (truncated or half Fourier imaging)
- or...generate and acquire multiple echos in one excitation
  (ETL = Echo Train Length)

If ETL>1 then rows in k-space are sampled at different echo times

- however, dephasing of T2 (SE) or T2* (GE) can not be neglected between two different echos, leading to:

\[ S(k_x, k_y) = H(k_x, k_y)S(k_x, k_y) \]

- this multiplication is a convolution in the spatial domain → blurring
Fast Sequences: TurboSE

SE sequence with 3-33 echos

- immediately after first echo apply new phase-encoding
- this selects a different line in k-space

Practice:

- divide k-space into $n$ segments
- within each excitation, sample one line in each segment

Example: T2-weighted imaging of a brain

- image size: 256x256
- TR=2500ms
- 4 segments
- acquisition time:

$$T_{A_{2D}} = \frac{1 \cdot 256 \cdot 2.5}{4} = 160 \text{ secs} < 3 \text{ minutes}$$
Half-Fourier Acquisition with Single-shot Turbo Echo (HASTE)

- all echos are generated within one excitation \( \Rightarrow \) ETL=\(N_{ph}\)
- and thus: TA = TR (gives heavily T2-encoded images)
- in fact: TA = time required to generate and sample ETL echos
- typically: 100 phase-encoding steps take 1s
- application: (motion-less) liver and lung imaging (single breath-hold)

Echo planar imaging (EPI)

- fastest sequence available today (acquisition time 100ms)
- GE sequence (no 180° pulse required, in contrast to TSE, HASTE)
- T2* dephasing limits number of echos measured above noise level
- maximum image size = 128x128
- requires high performance scanners to produce strong gradients
- used in functional MRI
Whole-body 1.5T Philips scanner (1.5 T, super-conducting magnet) standard diagnostic scanner

Open (C-shaped) scanner (0.2 T) helps with claustrophobia open space improves patient handling can be used for MRI-guided procedures SNR low and field homogeneity poor
Operation

head coil

body coil
Comparison: CT vs. MRI

Slice across a male prostate

MRI image: T2-weighted turboSE (TE=120ms, TR=6s)

CT image: over-estimates size of prostate (black outline) since MRI can distinguish between prostate tissue (white outline) and adjacent peripheral structures
MRI Tissue Characterization

(liver)  T1-weighted GE  T2-weighted (HASTE)

lesion

biliar cyst

hemangioma

early TE train  late TE train

hemangioma fades with TE (typical)
Functional MRI (fMRI)

Capitalizes on the oxygenation of cerebral bloodflow

- brain requires oxygen to function (via metabolic processes)
- active brain areas require more oxygen than inactive areas

Oxygen is transported in *hemoglobin* molecules

- in the arteries, hemoglobin carries 4 oxygen molecule $\rightarrow$ oxyhemoglobin
- in the capillaries, hemoglobin deliveres part of the oxygen to the neurons and carries only 2 oxygen molecules $\rightarrow$ deoxyhemoglobin

Distinguishing magnetic properties

- oxyhemoglobin is diamagnetic
- deoxyhemoglobin is paramagnetic $\rightarrow$ produces field inhomogeneities $\rightarrow$ these decrease T2* of blood and surrounding tissue

Knows as BOLD (Blood Oxygenation-Level Dependent effect)

- brain activation influences oxygen concentration in the blood
- this influences magnetic properties which can be measured with MRI
When brain cells are active, blood flow must increase to meet neuronal demands
  - in fact, blood flow over-compensates
    → too much oxygen in active brain areas
    → T2* is larger in active areas
  - GE (gradient echo) techniques are very sensitive to T2* variations
  - use EPI (echo planar imaging) for fastest acquisition speed

Typical experimental setup and acquisition protocol:
  - individual performs a certain experiment (for example, finger tapping) and MRI is taken
  - acquire two images: one at rest and one under task
  - subtract these two images to visualize the active areas

Many repetitions needed
  - signal difference is very small (2-5%) → low SNR
  - alternate periods at rest (30s) with active periods (30s) → 6 min total
  - take images every 2-10 s and process statistically
Get insight into:

- memory
- object recognition
- language
- visual cortex
- sensorimotor cortex
Capitalizes on the following facts:
- in different classes of tissue, diffusion of water molecules is different
- also, the diffusion is not isotropic → along fibers the diffusion is much greater than across fibers.

Acquisition and analysis
- diffusion-weighted images with diffusion gradients applied in different directions
- areas with high diffusion in that direction will appear dark
- pick a few diffusion directions for this
- perform eigenvector analysis → tensors
Diffusion Tensor Imaging (DTI)

Brain fibers can now be tracked using the local tensors
Diffusion Tensor Imaging (DTI)