DWI is now 20 years old...

45 minutes post MCA-0

H+

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MRI in Review:
Simple Steps to Cutting Edge Part I
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B0: High field at 7T
ΔB0: dB/dT < 3T/s

The main magnet B0.
Protons process 42.6 MHz / T.

The Gradient Coils alter B0 –
spatial localization or ΔB0.

The radiofrequency RF coil:
1. Creates B1 field with FM energy
2. Detects proton coherence in B0 -

Every electric field has a magnetic field...
Any moving magnetic field generates
an electrical current... Faraday’s Law

Coils in MRI

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Protons have a positive net charge.
This charge “spins” around the mass.

A moving charge is an electrical
“current” ...
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2. A moving magnetic field creates an electric field...

Spinning dipoles have local magnetic fields
which will align with external magnetic fields.
# protons aligned ~ Magnetic field strength (B0).
At 1.5Tesla, this is only 5 of every 1,000,000.
1 Tesla = 10,000 gauss.

Hydrogen nuclei are “protons”

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The proton spinning charge precesses or wobbles around the B0 axis... due to Angular Momentum.

The tip of the axis is related to the magnetic gyric ratio ($\gamma$).

The speed of the precession = Larmor Equation...

$$f = \left(\frac{\gamma}{2\pi}\right) B_0 \rightarrow 42.6 \text{ MHz@1T}$$

Know $B_0 = $ know frequency...

Know frequency = know $B_0$...

So, How Do We Get an MR Signal?

RF electric field creates a B1 magnetic field which “flips” protons away from B0... the “flip” is a “perturbation”...

1. As protons relax, they re-align along B0 by $T_1$.
2. They also lose phase coherence by $T_2^*$.  

Relaxation is the Process of Realignment

1. As protons relax, they re-align along B0 by $T_1$.
2. They also lose phase coherence by $T_2^*$.

Each proton has a unique $T_1$ and $T_2$

<table>
<thead>
<tr>
<th>Material</th>
<th>$T_1$ (msec)</th>
<th>$T_2^*$ (msec)</th>
</tr>
</thead>
<tbody>
<tr>
<td>White matter</td>
<td>390</td>
<td>90</td>
</tr>
<tr>
<td>Gray matter</td>
<td>520</td>
<td>100</td>
</tr>
<tr>
<td>CSF</td>
<td>2000</td>
<td>300</td>
</tr>
<tr>
<td>Muscle</td>
<td>600</td>
<td>40</td>
</tr>
<tr>
<td>Fat</td>
<td>100</td>
<td>90</td>
</tr>
<tr>
<td>Liver</td>
<td>270</td>
<td>50</td>
</tr>
<tr>
<td>Renal Med.</td>
<td>680</td>
<td>140</td>
</tr>
<tr>
<td>Renal Cor.</td>
<td>360</td>
<td>70</td>
</tr>
<tr>
<td>Blood</td>
<td>890</td>
<td>180</td>
</tr>
<tr>
<td>Tumor**</td>
<td>710</td>
<td>130</td>
</tr>
</tbody>
</table>

@ 1 Tesla
T1 is governed largely by protons interacting with membranes, lipids, cell walls, myelin... 

spin-lattice relaxation.

More macromolecules = large # of stationary protons = 
shorter T1 and shorter T2.

T2 of protons is dominated by interactions with other water 
protons, etc. T2 is affected by diffusion, perfusion, water 
content... spin-spin relaxation.

More water = large # mobile protons = less interaction = 
longer T1 and longer T2.

The spin-echo signal strength is related to T2 only...

We collect this signal in SE imaging.

The TE Determines the T2-weighting

Longer TE = more T2 phase decay and less signal. 
Shorter TE = less phase loss and more signal. 

"T2-weighted" image will have a long TE to show long T2 tissues

Proton relaxation is a process of re-aligning with B0.

1. As protons relax, they 
re-align along B0 by T1.

2. They also lose phase 
coherence by T2. 

The signal decay is the 
FID.

T1 is the recovery of magnetization 
along the longitudinal axis.

T2 is the decay of magnetization 
along the transverse axis.
Longitudinal Magnetization Recovers by T1 over Time

TR Dictates How Often RF is Applied

\[ M_z(T1, TR) \]

TR 600 ms

TR 2400 ms

The TR Determines the T1-weighting

The TR and TE Indicates the T1 and T2-Weighting

The TR and TE Indicates the T1 and T2-Weighting

IR for Morphology

Inversion Recovery Spin Echo

Fluid Attenuated Inversion Recovery (FLAIR)
Image Formation

Magnitude
Frequency
Phase

Readout gradient encodes protons according to precessional frequency along X...

Phase encoding gradient changes echo phase along Y at each frequency along X...

Gradient coils are built to create a magnetic field gradient (G/cm) along the x, y, and z axes to correspond to 3-D space...

The image orientation depends on which gradient is used for slice selection...

Current passing through gradient coils creates magnetic fields that add and subtract from B0 with no effect at isocenter.

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The MR Pulse Sequence.
MR signals are encoded with frequency Nx and phase Ny...

Slice-select 90° RF
Phase-encode
Slice-select 180° RF
Frequency-encode echo: k-space → image

Frequency-encode echo: k-space → image

TR
TE
Gx
Gy
Gz
Ny
T2
TE
TR
Nx
Ny
Gx
Gy
Gz

TE
TR
Gx
Gy
Gz
Ny
T2
TE
TR
Nx
Ny
Gx
Gy
Gz

Current in
back
isocenter
front
Current out
Subtracts from B0
No change
Adds to B0
1.4999 T
1.500 T
1.5001 T
63.7 MHz
63.9 MHz
64.1 MHz

\[ f = \frac{\gamma}{2\pi} (B0 + z Gz) \]
To excite protons in the head, we would need an excitation frequency of 64.1 MHz. Other protons at other positions and frequencies, would not be excited (wrong frequency).

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Changing Phase is Another Way of Measuring Frequency!
Protons alter phase with gradient pulses!

Variable strength

Gradient pulse strength

Phase change...

0 Time

Time

Time

RF coil

RF coil

1

2

3

3

2

1

RF coil

FT

Sequencing Issues and k-Space

Resolution

Bandwidth

TE

TR

FOVx, FOVy

Slice

In “Fast Spin Echo” we may collect up to 128 echoes...

A 90-180-180 Multiple Echo “pulse sequence”:

The echo signal strength is related to T2, TE, # echoes - # echoes is limited by T2

In FSE, we collect many echoes in a train...

A 90-180-180 double spin-echo “pulse sequence”:

The phase loss can be repeatedly reversed...

(as long as there is T2 signal left...)

The echo signal strength is related to T2 only...

The FID

The Spin-echo

Second echo

90 pulse

180 pulse

180 pulse

The phase change for each view is a frequency. FT of this frequency change gives projection along PE.

90 rf

180 rf

RF

Slice

Read

TE

TR

90 pulse

180 pulse
**Single-Shot FSE**

- **b=0**
- TR 6000
- TE 124
- average X, Y, Z
- FOV 24:
- 7.5/0 mm
- no gating

90 pulse 180 180 180 180

The FID

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**Parallel Imaging (PI)**

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**How Far Can We Go?**

32 Ch. 7T head coil and transmit coil

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**The "New" High Field Physics**

True 3D FSE - "CUBE", "SPACE", "VISTA"

3D FSE sequences with long FSE readout (>200 echoes).

Minimum echo spacing - fast acquisition - no artifacts.

Flip angle modulation during the readout keeps signal thru long train.

Provides best SNR, low SAR at effective TE.

Like CT, 3D allows for efficient reformats. Sequence can be easily modified for contrast.

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**Foundations of the Gradient Echo**

Apply a "gradient" for 1 ms... Reverse the gradient for 1 ms...

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**The Gradient Echo Pulse Sequence.**

MR signals are encoded with frequency $N_x$ and phase $N_y$...

- Slice-select RF
- phase-encode
- Frequency-encode echo: $k$-space $\rightarrow$ image
- time

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**3D FSE XETA**

Extended Echo Train Acquisition

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**3D FSE XETA - Reconstructions**

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High-Speed MRI - Today's Options

GRE EPI
TR 2000
TE 45
128x128

SE EPI
TR 2000
TE 80
128x128

What is Magnetic Susceptibility?
The magnetic susceptibility is the difference of the magnetic field across a sample. Each substance in a magnetic field alters that field.

Iron has a larger MS effect than water, e.g...

$B_{eff} = B_0 (1 - \gamma X)$

Bone - water
Air - water
Iron - water

Sequence Review

Observed T2*

α RF
si
4 msec
dephase
slicing selection gradients

REWIND or REFOCUS

2D spine MPGR

Long T2* remains, but is "in phase" with next shot.

At short TR’s, gradient-echo still dephasing beyond sampling time.

1. Transverse magnetization builds up from view to view.
2. Worse for long T2* fluids (CSF, etc). ARTIFACT.
Long $T_2^*$ crushed, by large gradient or by random RF.

**Contrast in GRE**

For long $T_E$ in GRE:
- Increase in $T_2$-wting....
- and MS artifacts

For long $T_R$:
- Increase in flip adds $T_1$-wting....

TR 200 Flip 30 MPGR