MRI Physics Course

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Magnetic Resonance Imaging (MRIA) 

Hemodynamics 

Imaging of flowing spins in MRI 

2D/3D Time of Flight Imaging Techniques 

2D/3D Phase Contrast Imaging Techniques 

CINE Phase Contrast Imaging Techniques 

MOTSA (Multiple Overlapping Thin Slab Acquisition)
Magnetic Resonance Angiography (MRA)

Signal intensity from fluid motion (flowing blood or CSF) in MRI depends upon several factors:

1.) flow velocity
2.) flow direction or pattern
3.) pulsatility (flow, acceleration, jerk)
4.) pulse sequence
A. Hemodynamics

Flowing blood components are subjected to friction forces from the vessel wall and from neighboring blood components.
Laminar is the most common type of flow distribution in the human body. Laminar flow is characterized by a parabolic velocity profile where the velocities in the center of the vessel are greater than those at the vessel wall (zero velocity at the wall).
High velocity fluid motion through stenotic vessels and arterial branches (e.g. carotid bifurcation) produces complex flow patterns including flow vortices and turbulent flow:

- **Flow vortices (recirculation)**
  
  A flow vortex occurs due to rapid deceleration of the fluid when flowing through a small orifice or at the site of an arterial branch.
MRA (continued)

Vortex flow is slowly swirling (usually counter-current) pools of fluid separated from the high velocity component of flow.
MRA (continued)

Turbulent flow is described as chaotic with randomly fluctuating velocities. Unlike laminar flow, the velocity profile of turbulent flow tends to be flatter.
In MRA, flow vortices and turbulent flow cause increased dephasing of spin coherence due to flow in many directions, acceleration and higher order motions. Increased dephasing can contribute to signal void. In MRA, turbulent flow just downstream of stenosis can increase signal void and make the degree of stenosis appear to be larger than it actually is.
Turbulent flow is characterized by a fluid’s Reynolds number (Re) which depends upon the following factors:

- \( \rho \) = fluid density \((\text{g/cm}^3)\)
- \( D \) = vessel diameter \((\text{cm})\)
- \( V \) = flow velocity \((\text{cm} / \text{sec})\)
- \( \eta \) = fluid viscosity \((\text{N sec/m}^2)\)

[remember that \( T_1 \) and \( T_2 \) also depend on viscosity]

\[
Re = \left( \rho DV \right) / \eta
\]

\( Re < 2000 \) usually represents laminar flow
\( Re > 2000 \) usually represents turbulent flow
Pulsatile Flow:
The magnitude and phase of the signal from pulsatile flow is a complex function of time. These temporal changes can cause spatial misregistration of pulsatile flow in images.

Venous flow is much less pulsatile than arterial flow. Flow velocities in the human body under normal conditions range from a few mm/sec up to 180 cm/sec. Velocity decreases as distance from the heart and cross sectional area of vessels increase.
<table>
<thead>
<tr>
<th>Intracranial Vessel Peak Velocities</th>
<th>Peripheral Vessel Peak Velocities</th>
</tr>
</thead>
<tbody>
<tr>
<td>Vessel</td>
<td>Peak Velocity (cm/sec)</td>
</tr>
<tr>
<td>Middle cerebral artery</td>
<td>62 ± 12</td>
</tr>
<tr>
<td>Anterior cerebral artery</td>
<td>52 ± 12</td>
</tr>
<tr>
<td>Posterior cerebral artery</td>
<td>42 ± 10</td>
</tr>
<tr>
<td>Internal carotid siphon</td>
<td>54 ± 13</td>
</tr>
<tr>
<td>Vertebral artery</td>
<td>36 ± 9</td>
</tr>
<tr>
<td>Basilar artery</td>
<td>42 ± 10</td>
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<tr>
<td></td>
<td></td>
</tr>
<tr>
<td>average venous flow</td>
<td>2 - 5 cm / sec</td>
</tr>
<tr>
<td>average arterial flow</td>
<td>10 - 20 cm / sec</td>
</tr>
</tbody>
</table>
Flow effects of moving spins (flowing blood):

1.) Phase changes: In a gradient magnetic field, spins accumulate phase depending on their position.

When spins move into a region of higher magnetic field, their phase increases with time. When spins move into a region of lower magnetic field, their phase retards with time.
Spins flowing at constant velocity in a gradient magnetic field will accumulate phase (relative to static spins) based on

- Direction of motion
- Speed
- Gradient amplitude
- Gradient duration
For a bipolar gradient:
Flow effects of moving spins (flowing blood):

2.) Time of Flight effects:

Saturation differences: Flow perpendicular to an imaged slice can result in spins entering the slice that have a different history of RF excitation than static spins in the slice. Thus, spins flowing into a slice may be less saturated than static spins resulting in angiographic contrast differences.
MRA (continued)

B. Imaging flowing spins in MRI

Signal loss (or flow voids) from flowing spins in MRI is primarily attributed to:


2.) Loss of phase coherence due to the distribution of velocities in the vessel.
MRA (continued)

Loss of phase coherence (phase dispersion) in spins having constant velocities can be recovered with addition of gradient waveform lobes. Known as:

- first order gradient moment nulling
- flow compensation
- gradient motion rephasing

First order gradient moment nulling employs gradient pulses with three lobes. The first and third lobes are equal in magnitude and duration but the middle lobe is reversed and twice the magnitude of the others. The effect of this gradient waveform is to re-establish phase coherence for stationary spins and spins flowing at constant velocity.
slice select gradient

phase of stationary tissues

phase of constant velocity spins
Due to the additional lobes on the gradient waveforms, the minimum TE is increased.

Spins experiencing higher orders of motion such as constant acceleration are not compensated by this technique as shown in the following slide:
TE

slice select gradient

phase of stationary tissues

phase of constant acceleration spins
Blood flow is pulsatile!:
Velocity: rate of change of position
Acceleration: rate of change of velocity
Jerk: rate of change of acceleration
- It is possible to rephase constant velocity and constant acceleration by adding an additional lobe to the flow compensating gradients. This is called 2nd order gradient motion rephasing or 2nd order gradient moment nulling which compensates for constant acceleration. Changes in the rate of acceleration are described as jerk. Jerk can be compensated for by 3rd order gradient moment nulling or 3rd order gradient motion rephasing with the addition of yet another lobe to the flow compensating gradients.

- Most flow compensation techniques currently in use only compensate for 1st order effects from constant velocity flow.
Time of Flight (TOF) techniques

Time of flight techniques use signal amplitudes of flowing spins to enhance contrast between stationary tissues and vasculature. There are two time of flight effects in MRI to consider:

1.) Black blood (Spin Echo) techniques: out-flow of excited spins from the slice of interest create signal loss
In conventional Spin Echo imaging techniques flow voids are primarily caused by the time of flight effect. The flowing spins excited in a slice of interest must experience both the 90° RF pulse and the 180° refocusing RF pulse to produce an echo signal. For higher flow velocities (> 25 cm/sec) most of the flowing spins have exited the slice of interest prior to application of the 180° RF pulse.
The degree of intraluminal signal loss in black blood techniques is influenced by flow velocity, slice thickness, echo time and the angle between the blood vessel and the slice.
MRA (continued)

2.) Bright blood techniques: in-flow of fresh unmagnetized spins or “flow related enhancement” resulting in increased signal intensity

Repetition times (TR) in conventional MRI techniques do not allow full recovery of the longitudinal magnetization for tissues having $T_1 > TR$. Stationary spins become saturated and contribute reduced signal when repeatedly flipped into the transverse plane by 90° RF pulses or $\alpha$ pulses.
Unmagnetized flowing spins continually replace saturated flowing spins in the imaging volume and produce very high signal intensity when flipped into the transverse plane by an RF pulse.

Signal intensity from flowing spins increases with velocity.
Flow related enhancement is prominent in gradient recalled echo techniques where TR and TE are very short. Short TRs provide stationary spin saturation and short TEs reduce the amount of time for motion related phase dispersion to occur.
Sat Bands:

Pre-saturation bands are 90° RF pulses applied outside the field of view. The resulting transverse magnetization is then nutated to 180° by a 90° slice selective RF pulse. Pre-saturation selectively suppresses signal from venous/arterial flow when applied downstream/upstream of arterial flow. For example, inferior pre-saturation bands will suppress signal from arterial flow in the head. Pre-saturation techniques reduce phase ghosting artifacts and decrease intraluminal signal.
MRA (continued)

C. Two-dimensional Time of Flight Angiography (2DTOF)

2D time of flight sequences are gradient recalled echo techniques in which flow related enhancement (inflow) produces contrast between stationary and flowing spins.

Flow compensation (GMR: gradient motion rephasing) is generally used to reduce signal loss from flowing spins caused by phase dispersion.
MRA (continued)

Signal from stationary tissues is suppressed by the frequent (short TR) slice selective RF excitations and by magnetization transfer preparation.

Multislice studies are acquired one slice at a time.

Pre-saturation bands can be used to suppress signal from venous or arterial flow.

Flow related enhancement is maximized when the direction of flow is perpendicular to the slice.
MRA (continued)

At MCG:

2D TOF MRA is used primarily for:

Cervical MRA

MRV
2D TOF MRV: collapse image
MRV

2D TOF MRV: source image
D. Three-dimensional Time of Flight Angiography (3DTOF)

3D time of flight sequences are three dimensional gradient recalled echo techniques in which slices are encoded using a slice encoding gradient ensemble and flow related enhancements (inflow) produces contrast between stationary and flowing spins.

Partial saturation of stationary spins reduces signal intensity.
- Fresh, unsaturated flowing spins produce high signal especially in the entry slices.

- Flow related enhancement is most obvious in the slice that blood first enters the imaging volume and is commonly called entry slice phenomenon.

  The entire stack of slices (slab) experiences every RF excitation

  Slow slowing spins may become saturated before exiting the stack

  3DTOF techniques produce reduced contrast in the downstream side of the stack.
Flow compensation will recover phase dispersion caused by constant velocity motion but not higher order motion (pulsatile flow).
3D Time of Flight Angiography Pulse Sequence

RF

flow comp

slice select

slab select

slice encode

phase encode

flow comp

frequency encode

TE

spoiler

rewinder
MRA (continued)
MRA (continued)

3D TOF MRA: source image  1.5T
MRA (continued)

3D TOF MRA: collapse image 3T
3D TOF MRA: projection image 3T
1 of 19 projections
MRA (continued)

3D TOF MRA: source image 3T
MRA (continued)

Flow Related Enhancement

3D TOF MRA
TR:20  TE:3.4  Flip:15°
160 images in 3 minutes

Flow Voids

T2 FLAIR PROP
FSE
TR:9000  TE:148
MRA (continued)

Flow Related Enhancement
3D TOF MRA
TR:20  TE:3.4  Flip:15°
160 images in 3 minutes

Flow Voids
T2 FLAIR PROP
FSE
TR:9000  TE:148
3.0T SNR Compared With 1.5T

3.0T MR Angiography Exceeds DSA Resolution

1024 Time of Flight
No Contrast

1024x608
16 x 12cm FOV
60-0.8mm slices

8-Channel head coil

Images courtesy of Lawrence Tanenbaum, M.D. Edison Imaging, New Jersey
3.0T SNR Compared With 1.5T

Note how well the lenticular striates are demonstrated

Images courtesy of Lawrence Tanenbaum, M.D. Edison Imaging, New Jersey

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At MCG:

3D TOF MRA is used primarily for:

- Brain MRA
E. Two-dimensional Phase Contrast Angiography (2DPC)

- 2D phase contrast sequences are gradient recalled echo techniques in which velocity induced phase shifts produce contrast between stationary and flowing spins.

- Good technique for slow (venous flow) in small vessels (e.g. MRV).

- Bipolar gradients are used to encode spin velocity as a phase change.
Flow induced phase shift is directly proportional to flow velocity.

Static tissues will have no net phase accumulation in the presence of the bipolar gradients.
MRA (continued)

2D techniques having acquisition times of approximately 1 - 3 minutes can be repeated using several values of the bipolar Velocity Encoding gradients (VENC) to emphasize slow flow, fast flow, venous flow or arterial flow.

In angiographic studies the VENC gradients are specified by the fastest flow encoding capability (cm/sec) rather than Gauss/cm.

VENC must be selected for the fastest anticipated flow to prevent phase accumulation of more than +180° or -180° (which becomes ambiguous).
MRA (continued)

At MCG:

Phase contrast is used primarily as 2D cine phase contrast for CSF flow.
MRA (continued)

F. Three-dimensional Phase Contrast Angiography (3DPC)

-A 3D technique where slices are encoded using an ensemble of slice encode gradients.

-Available but not used at MCG.
3D Phase Contrast Angiography Pulse Sequence

RF

slice select

flow encode

phase encode

frequency encode

spoiler

rewinder

TE
G. Resultant Angiographic Images

The results of angiographic studies can be displayed variously:

Magnitude images: Conventional gradient recalled images (usually with flow compensation or GMR) of each slice or slab. Flow enhancement is primarily due to Time of Flight effects.
Maximum Intensity Projection Images (MIP):

Typically 19 or 37 MIP images whose projections are rotated around the slice select axis of a slab or stack of slices in 20 degree or 10 degree steps.
MRA (continued)

3D TOF Source Images