Introduction to Medical Imaging

Lecture 12: 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 ω_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 ω_0) characterized by amplitude A and phase shift θ_s → 2 knowns
- what we want is the density of nx·ny·nz individual voxels → nx·ny·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

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 ω_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 ω_s

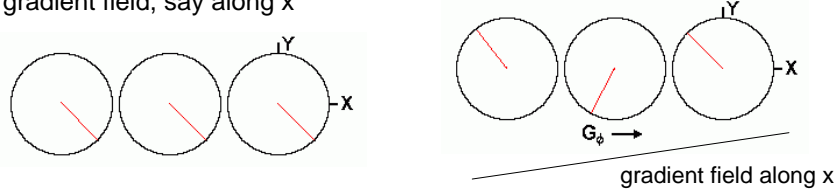gradient field

Step 2: RF pulse at ω_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)
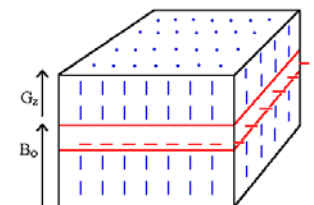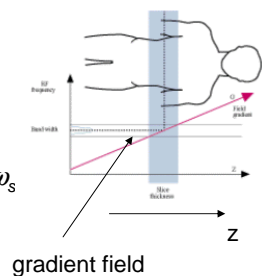
The Phase Encoding Concept

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

gradient field 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
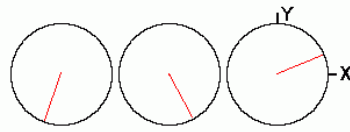
- we need a method to measure this phase shift
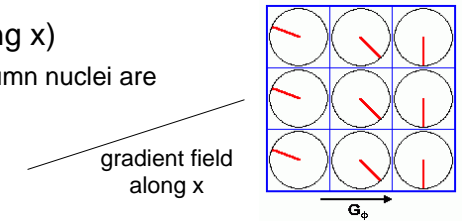- we also still need a method to distinguish different rows *y*

Two-Dimensional Encoding

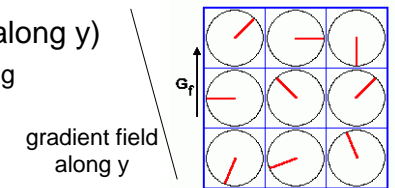After slice selection all nuclei in the slice spin at the same rate
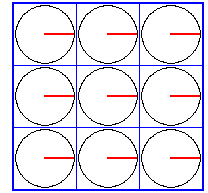
Apply a phase encoding (along x)
- after gradient release, the column nuclei are distinguished by phase

gradient field along x

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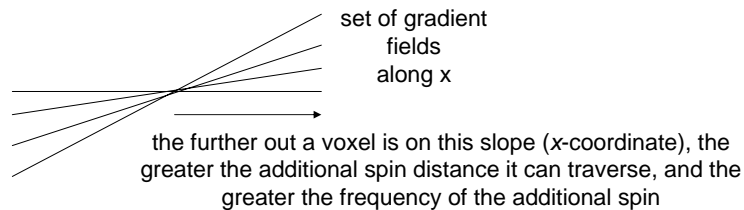
gradient field along y

Decoding

We could measure the RF signal and perform a Fourier transform
- this would give us *ny* frequencies
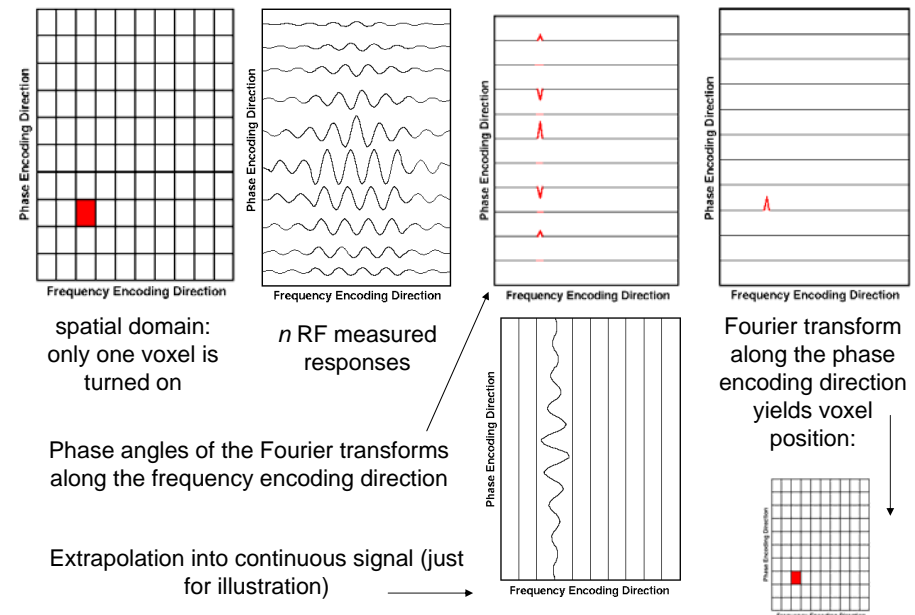- however, we would like to measure *nx·ny* voxels

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
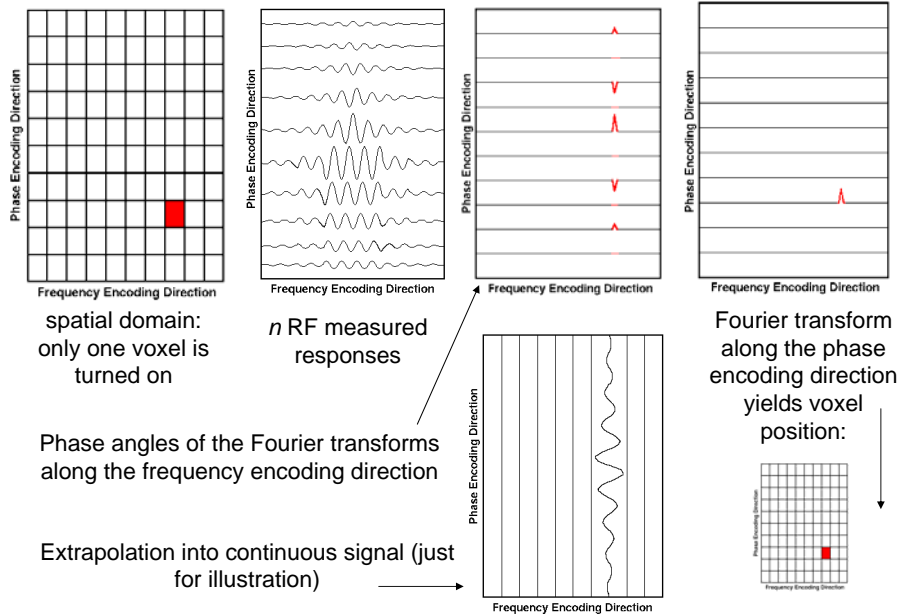

set of gradient fields along x

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

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 2

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 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)

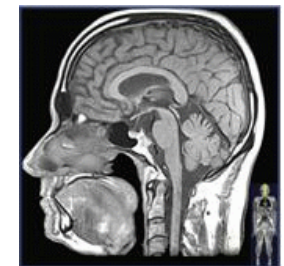Fourier transform along the phase encoding direction yields voxel position:

Image Generation

The spectra have amplitudes as well and an image can be generated
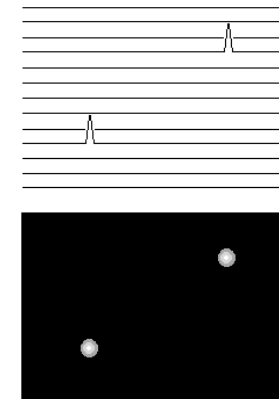
or more complex:

Gradient Coils

Field gradients, generated by coils, are essential in MRI
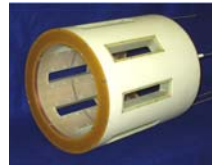
x-direction:

y-direction:

z-direction:

Examples:

surface coils

volume
(quadrature) coil

head volume coil

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:

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

$$\Delta z = \frac{BW}{\gamma G_z}$$

$$BW = \Delta \omega$$

- the Larmor frequency of a slice is then:

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

- hence, a slice of thickness Δz 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 Δz:

- technical and safety limits on gradient strength G_z
- a very thin slice has too low SNR
- typically: Δz = 2 mm (B_0=1.5T), Δz = 3 mm (B_0=1T)

$$\Delta z = \frac{\Delta \omega}{\gamma G_z} = \frac{BW}{\gamma G_z}$$
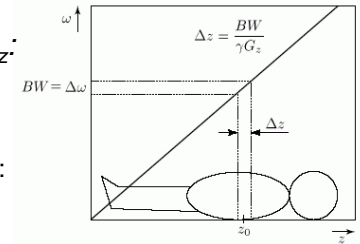
Slice Selection: Example

Dynamic cardiac sequence

- four cardiac chambers together with the heart valves in a plane parallel to the cardiac axis

Sc10/2
B-FFE / M
Gd 034 ms

L

L 47
A 37
F 49

W 1649
L 825

Position Encoding: Spin Differentiation

The slice selection RF pulse puts all nuclei within Δz into a similar synchronized transverse spin (resonance)

z

- 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 ω_0, in a synchronized fashion
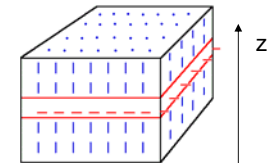
Let's assume, for now, that we only have one x-beam of voxels
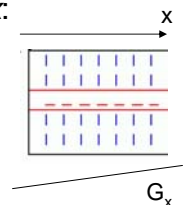
For encoding, we apply a field gradient G_x along x:

x

- this changes the transverse component frequency to:

$$\omega(x) = \gamma G_x x$$

- in complex notation:

$$M_x(t) = M_x(0)e^{-i\gamma G_x x t}$$

G_x

Position Encoding: Signal Collection

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 $k_x = \dfrac{\gamma}{2\pi} G_x t$ we get: $s(t) = S(k_x) = \int_{x=-\infty}^{+\infty} \rho(x)\, e^{-i2\pi k_x x} dx$
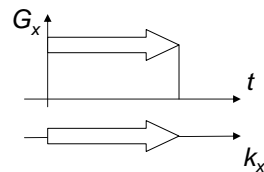
MRI measures: $f(x)=\rho(x)$

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)

The K-Theorem (1)

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 $k_x = \dfrac{\gamma}{2\pi} G_x t \quad k_y = \dfrac{\gamma}{2\pi} G_y t$)

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

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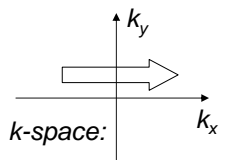
The K-Theorem (2)

Generalizing further into 3D:

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

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

$$s(t) = \int_{-\infty}^{+\infty}\int_{-\infty}^{+\infty}\int_{-\infty}^{+\infty} \rho(x,y,z) g(t,T_1)\, \mathrm{h}(t,T_2) e^{-i\gamma(\vec{G}\cdot r)t} dx\, dy\, dz$$
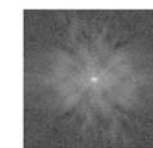
MRI seeks techniques where the returned vales at (*x,y,z*) all have *the same* relaxation effects

- 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* (ρ) image (but this can be obtained as well)
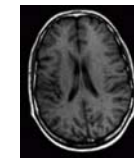

The K-Theorem (3)

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 *k-space*
- an inverse FT will produce the desired image

2D Fourier transform

k-space image   spatial 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

running at different speeds

runners transposed
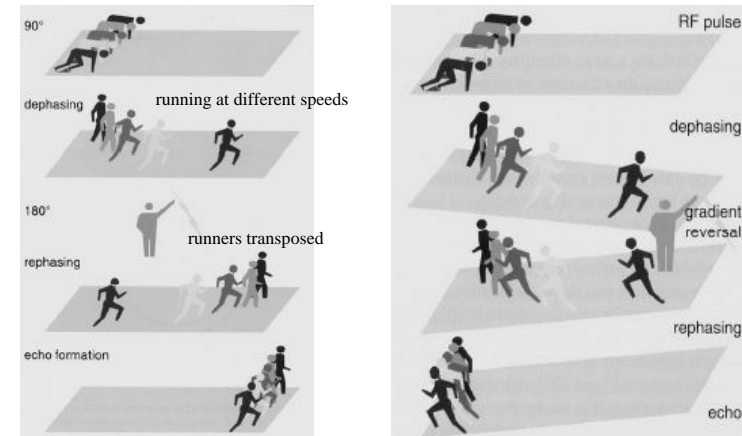
180° RF pulse   Gradient reversal

from: Frits Thorsen (U Bergen)

Rephasing: 180 RF Pulse

dephase time= rephase time

dephasing

rephasing

acquisition

TE = Time to Echo

acquired signal, reduced by T_2 effects
degree of reduction depends on choice of TE
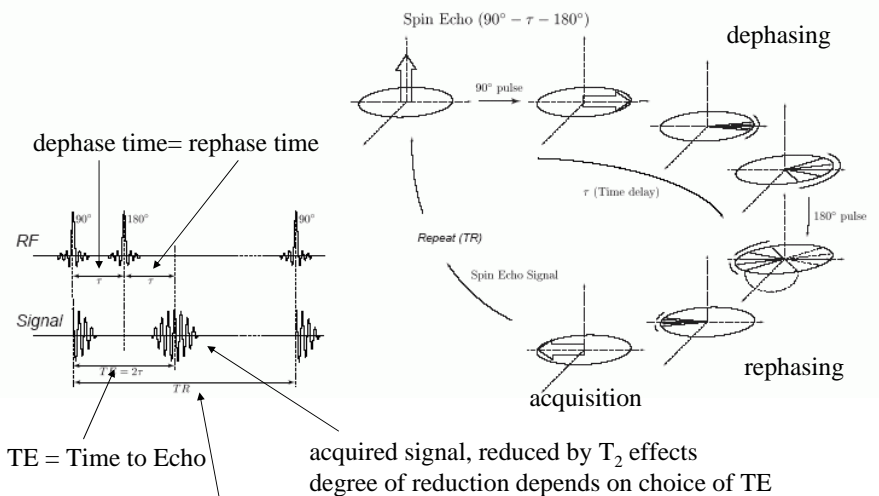
TR = Repetition Time (can also be chosen for different effects)
large TR ensure that all protons are aligned with the z-axis before the next RF pulse

Rephasing: Gradient Reversal

Opposite field gradients of same length and strength
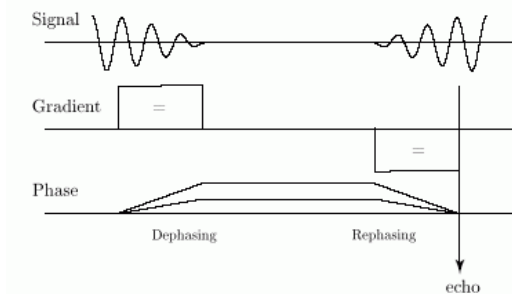
The Art of Choosing TR and TE

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

T1

T2
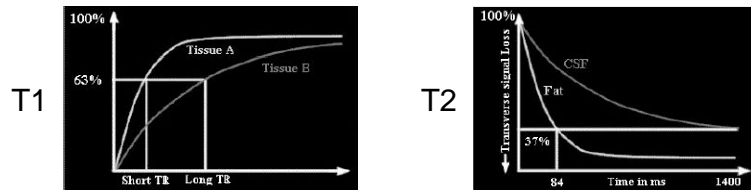
Examples

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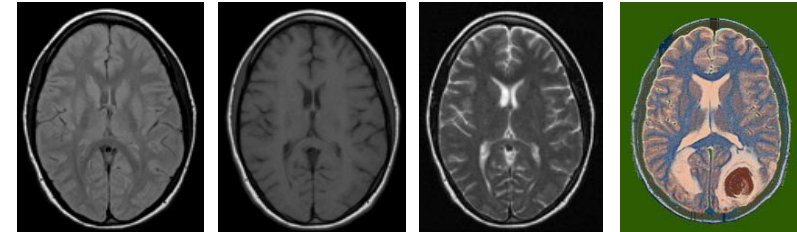
proton density   T1-weighted   T2-weighted   T1/T2 fused

T2 Weighting

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

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

Brain

CSF

M_z M_{xy} Echo (measurement)

M_{zo}

90° RF pulse

TR   TE
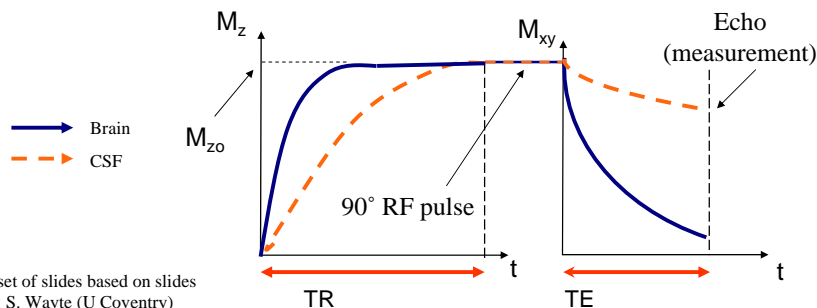
next set of slides based on slides from: S. Wayte (U Coventry)

T2 Weighting

z B_0 z B_0 Brain CSF

M_{xy}

y y

x x

Flip Magnetization   Long TE Time

Long TR Time

T2 Weighting: Examples

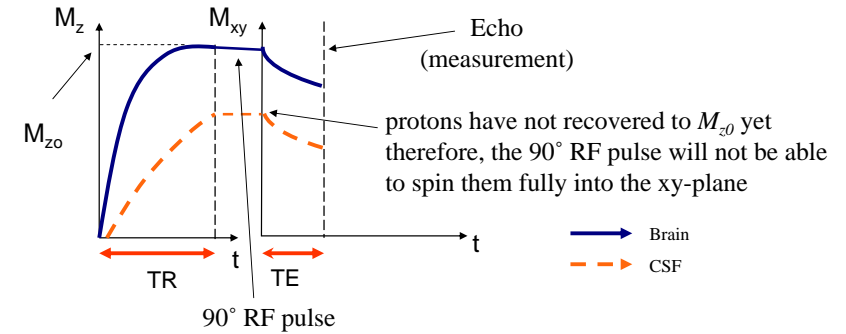TR = 3070ms   TE = 92ms
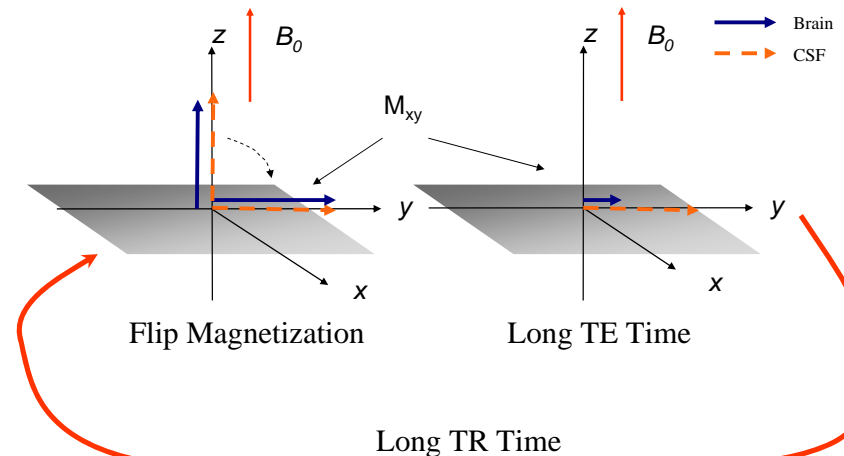
TR = 4000ms
TE = 132ms
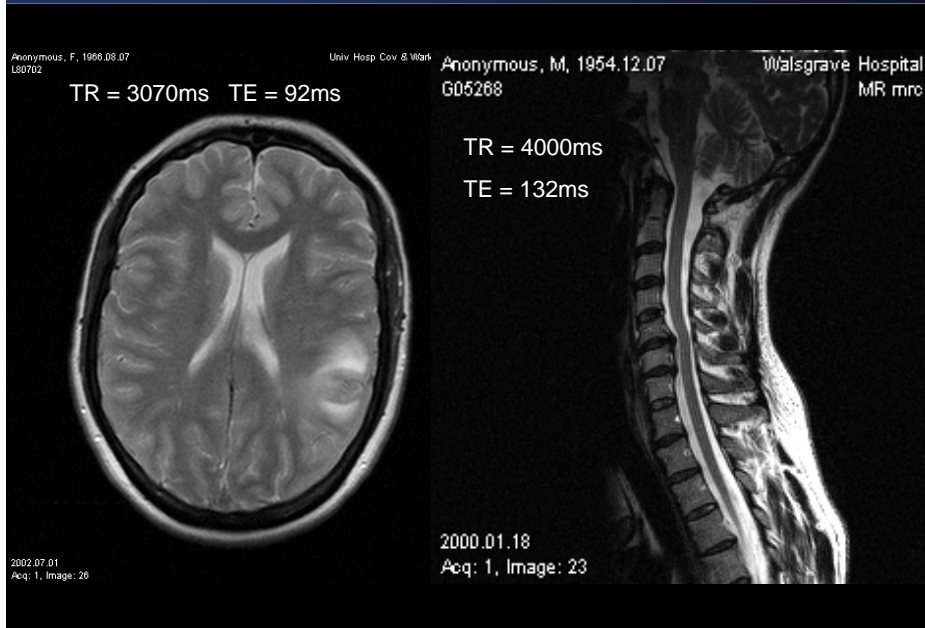
T1 Weighting

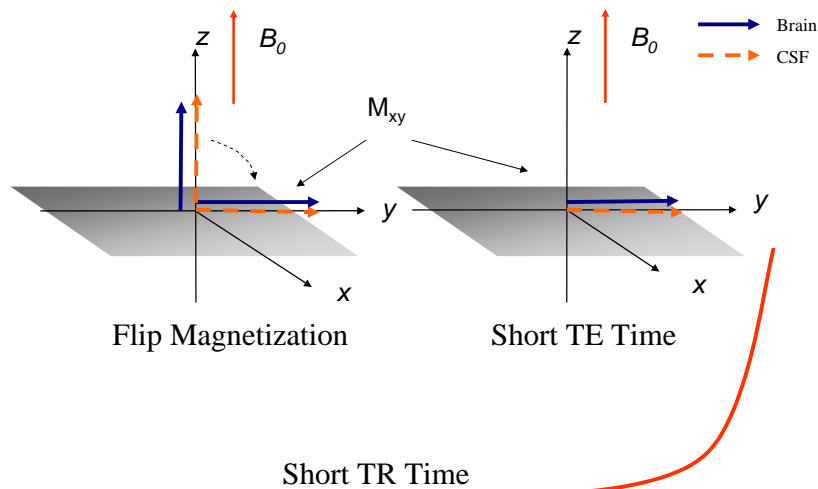Short TR
- to bring out differences in T1 between tissues

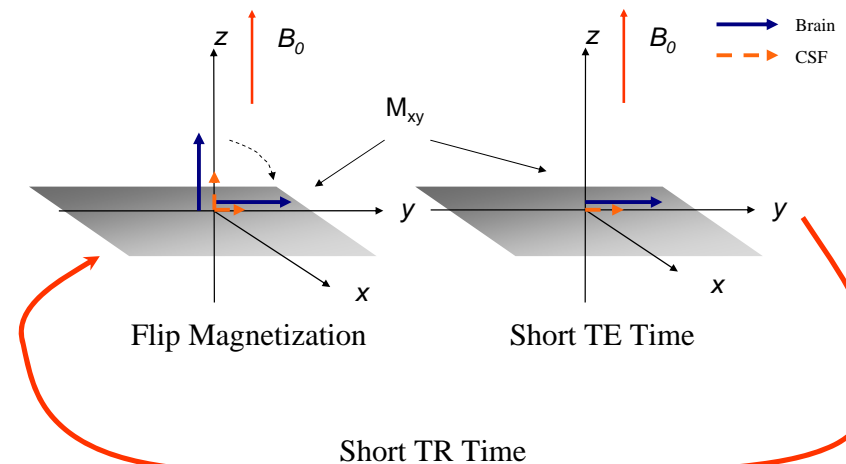Short TE
- so differences in T2 do not have time to appear

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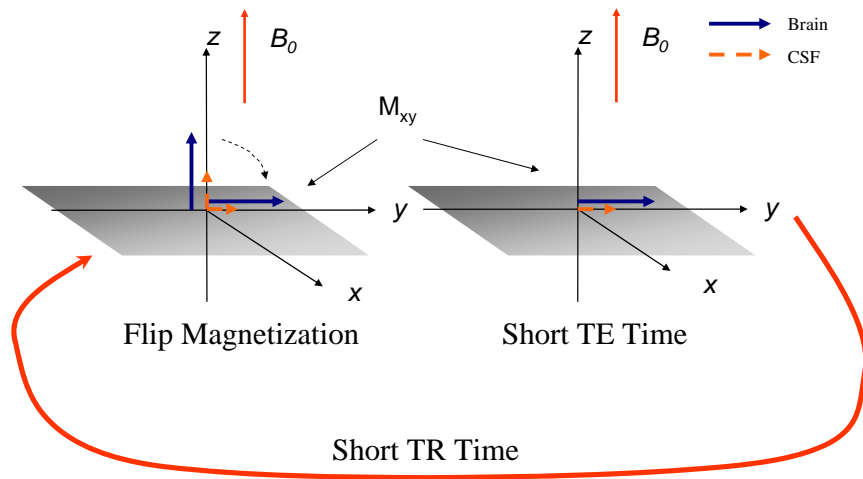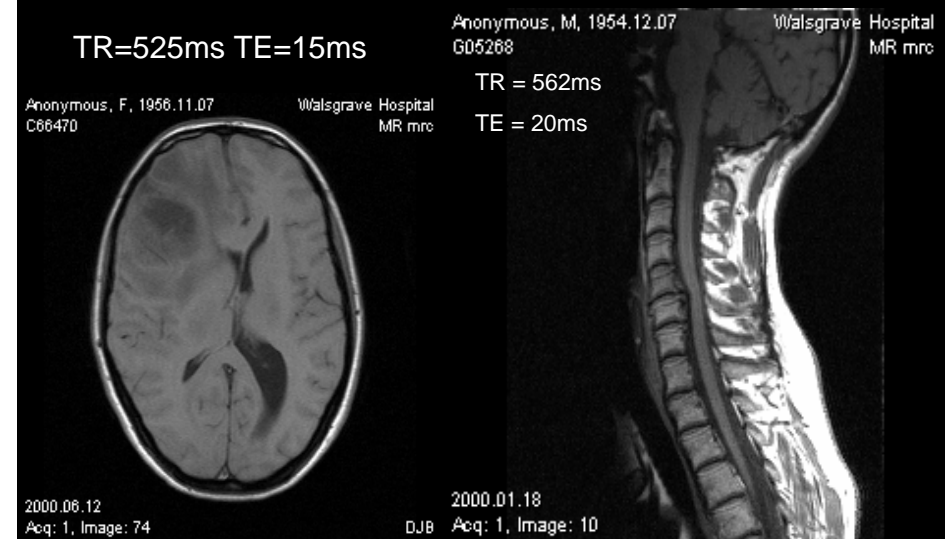

Echo (measurement)

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

Brain
CSF

90° RF pulse

T1 Weighting: First Time Around

Flip Magnetization   Short TE Time

Short TR Time

T1 Weighting: Second Time Around

Flip Magnetization   Short TE Time

Short TR Time

z B_0

z B_0

Brain
CSF

M_{xy}

y

y

x

x

Flip Magnetization

Short TE Time

Short TR Time

TR=525ms TE=15ms

TR = 562ms

TE = 20ms

Substance	T1 Weighted	T2 Weighted
Water/Vitrious/CSF	black	light grey or white
Fat	white	light gey
Muscle	grey	grey
Air	black	black
Fatty bone marrow	white	Light grey
Brain white matter	light grey	grey
Brain grey matter	grey	Very light grey

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

T2 weighted
  long TR = 1500ms (3xT1 max)
  long TE > 80ms

TR = 562ms

TE = 20ms

TR = 4000ms

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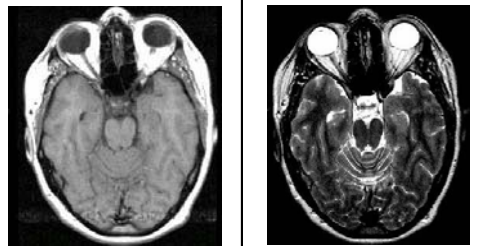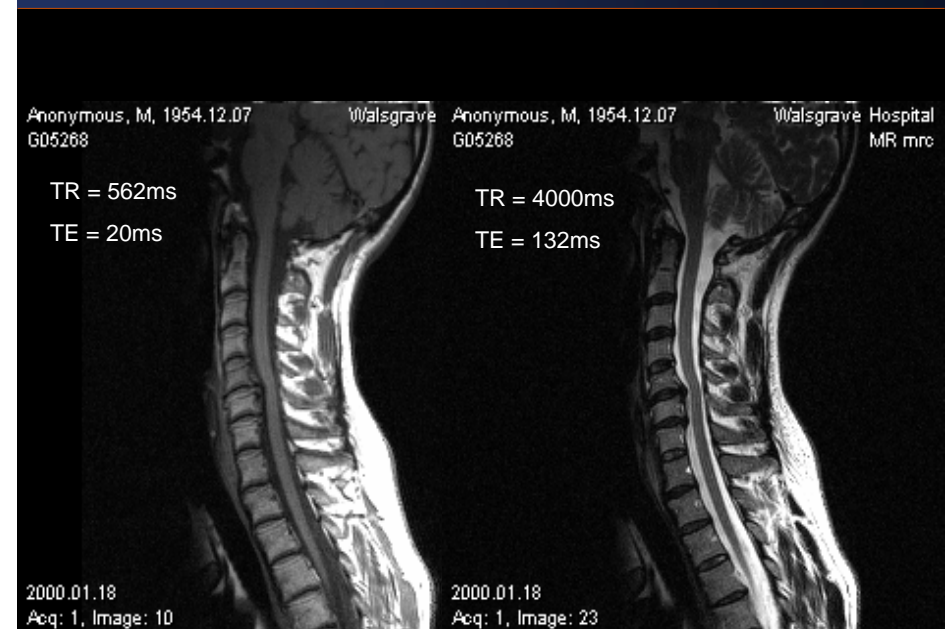
T1 vs. T2 Weighting (3)

TR=525ms TE=15ms
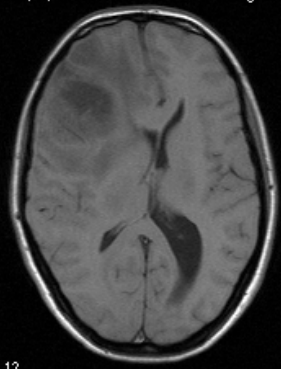
TR=2500ms TE=85ms
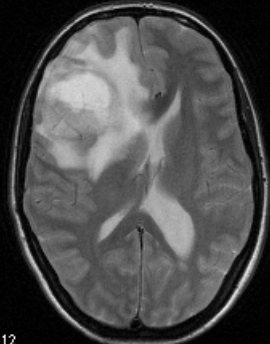
Proton Density Weighting

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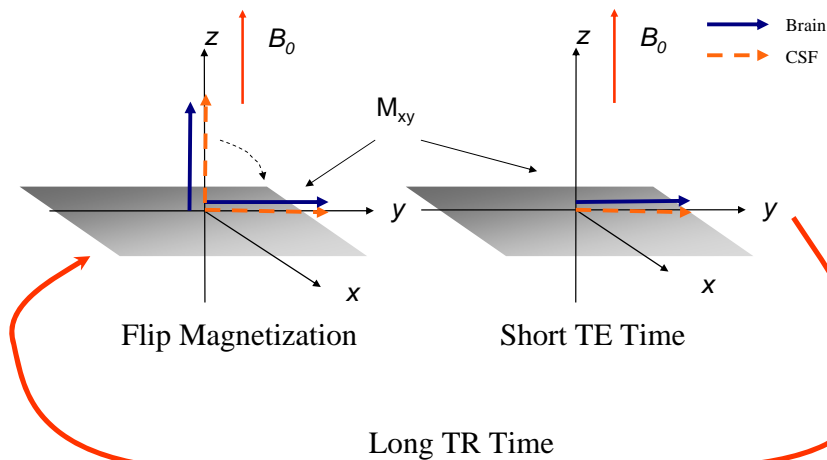- to give all tissue time to recover and thus eliminate T1 weighting

Short TE
- to give no T2 weighting

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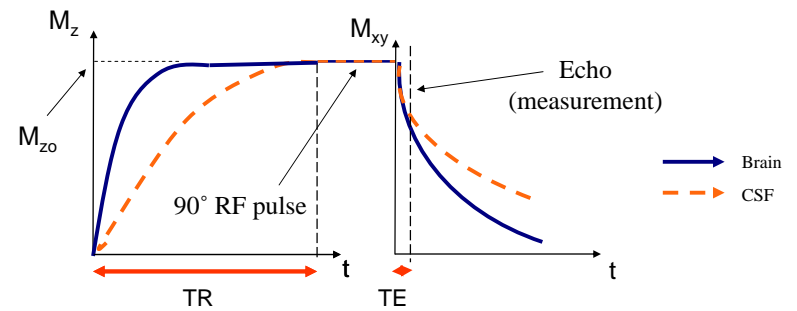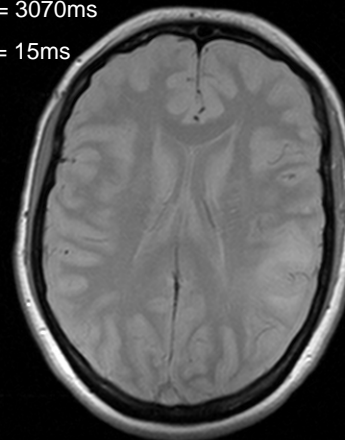

Proton Density Weighting

Flip Magnetization

Short TE Time

Long TR Time

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
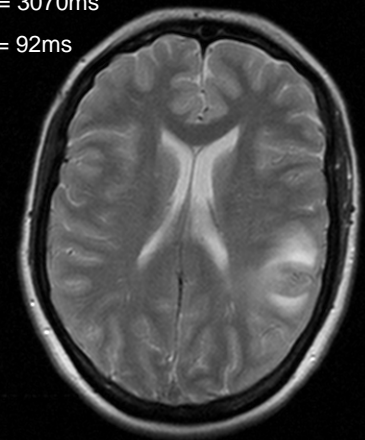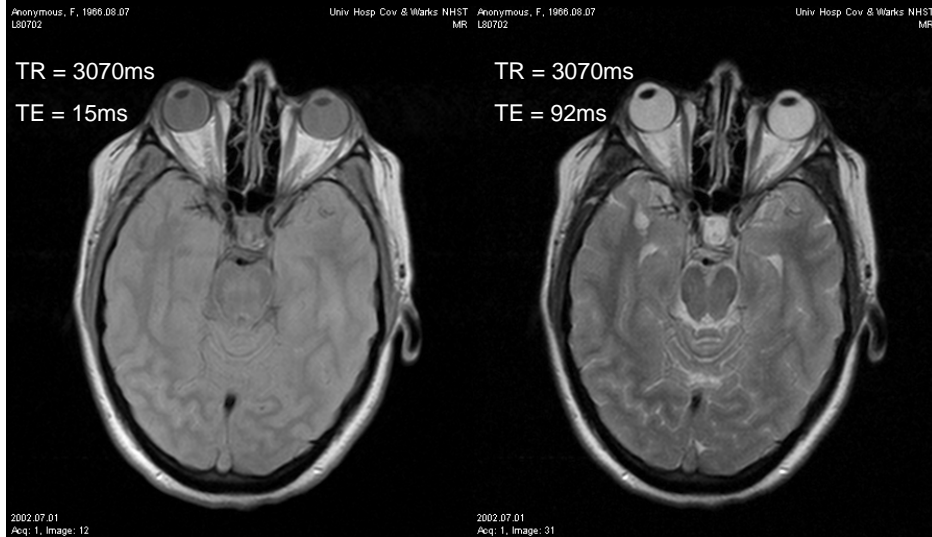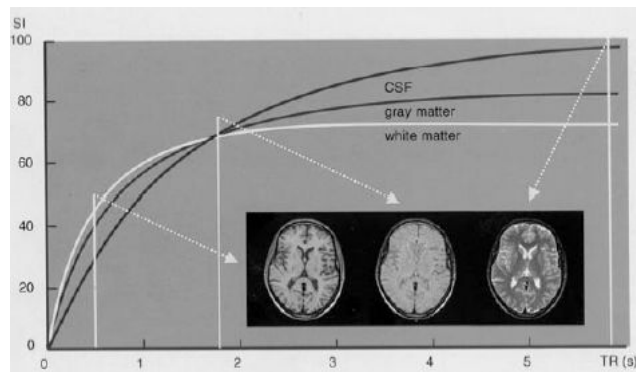
Proton Density vs. T1

Knee with Pigmented Villonodular Synovitis (PVNS)
- example shows that PD and T1 weighting creates contrast for complementary features
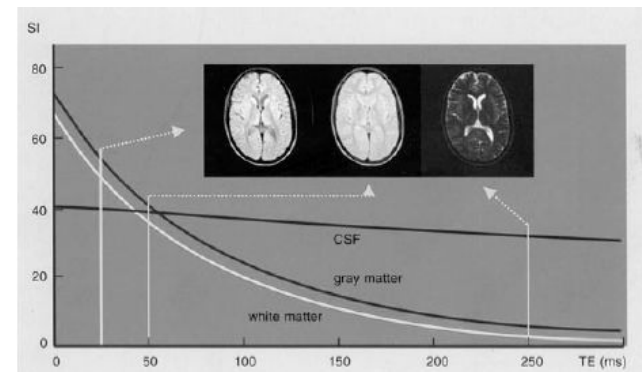

Contrast Dependence on TR

Use TR to maximize contrast among tissues of interest

from: Frits Thorsen (U Bergen)

Contrast Dependence on TE

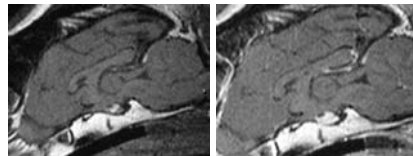Use TE to maximize contrast among tissues of interest

from: Frits Thorsen (U Bergen)

Other Techniques

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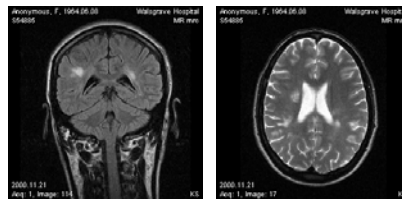

without   with Gadolinium

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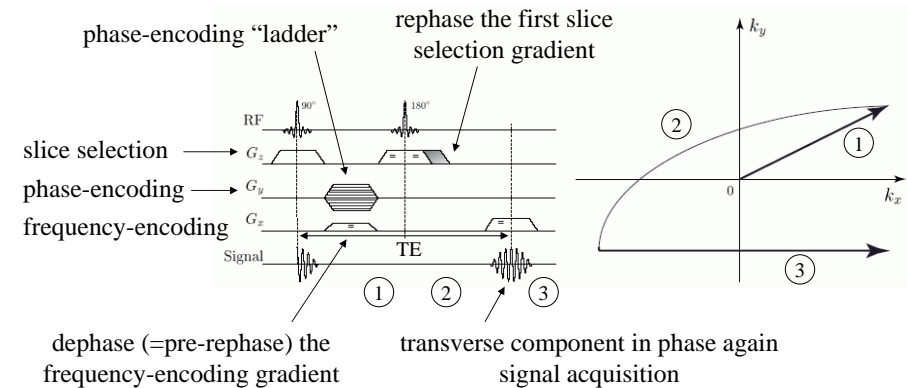- tends to be slower than regular T1

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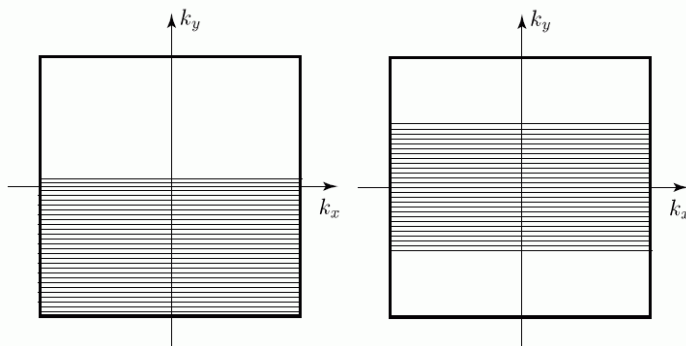

And many more…

Sequences: Spin-Echo

phase-encoding "ladder"

rephase the first slice selection gradient

dephase (=pre-rephase) the frequency-encoding gradient

transverse component in phase again signal acquisition

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

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

$$k_y = \frac{\gamma}{2\pi} mg_y T_{phase-encoding}$$

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 ρ 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 → 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
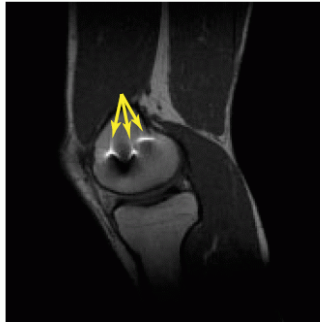- instead perform faster earlier acquisition → measure T_1 effects

with small flip angles short TRs return a larger signal
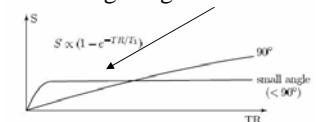
Comparison: 180° RF Pulse vs. Gradient Echo

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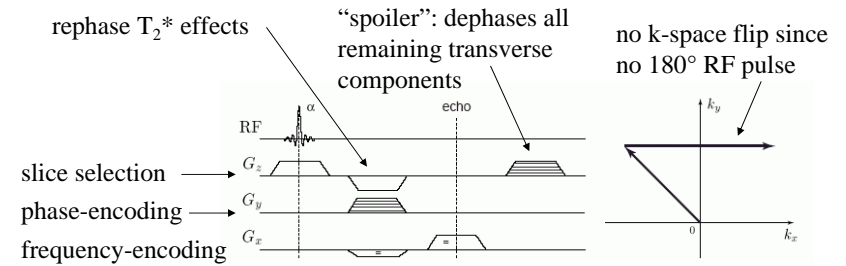- example: ferromagnetic particles in a knee joint

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° 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
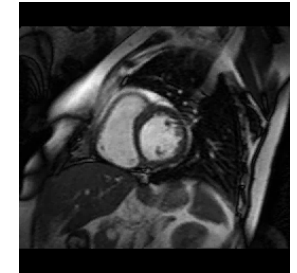
Three-Dimensional Imaging

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

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

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
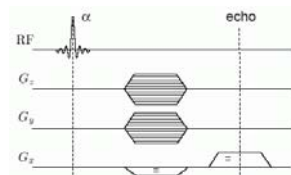
Acquisition and Reconstruction Time

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: $TA_{2D} = N_{ph} N_{ac} TR$
- 3D pulse sequences: $TA_{3D} = N_{ph} N_{sl} N_{ac} TR$

- 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)

3D Clinical Imaging Example: Brain

Coronal cut   Sagittal cut   Axial cut

Assume a 32 x 256 x 256 volume:

- N_{ac}=1, N_{sl}=32, N_{ph}=256
- with Spin-Echo, TR=2000 ms → TA_{3D} = 4 hours (infeasible)
- with Gradient-Echo, TR=40ms → TA_{3D} = 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 excitiation
  (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
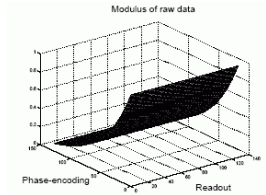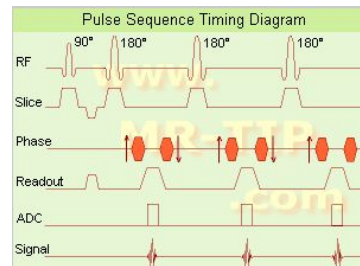
Fast Sequences: TurboSE

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:

$$TA_{2D} = \frac{1 \cdot 256 \cdot 2.5}{4} = 160 \text{ secs} < 3 \text{ minutes}$$

Fast Sequences: HASTE and EPI

Half-Fourier Acquisition with Single-shot Turbo Echo (HASTE)

- all echos are generated within one excitation → 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 180⁰ 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
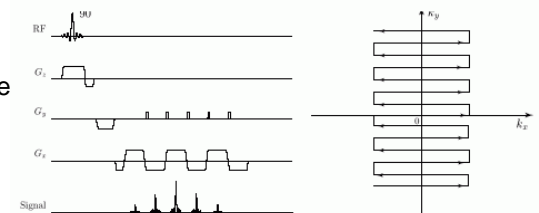
Clinical Equipment

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
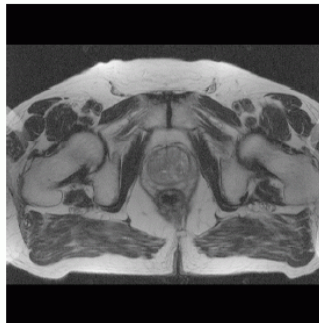
Operation

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 betyween prostate tissue (white outline) and adjacent peripheral structures

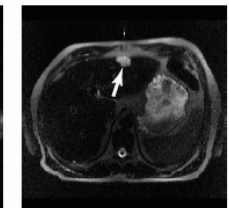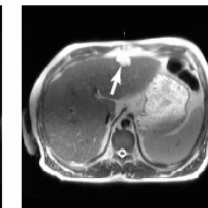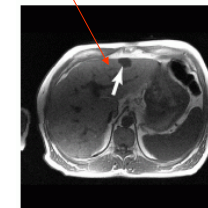## MRI Tissue Characterization
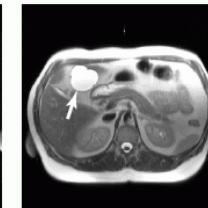
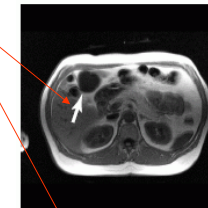(liver)   T1-weighted GE   T2-weighted (HASTE)

early TE train   late TE train

lesion

biliar cyst

hemangioma

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
  → oxyhemoglobin
- in the capillaries, hemoglobin deliveres part of the oxgen to the neurons and carries only 2 oxygen molecules → deoxyhemoglobin

Distinguishing magnetic properties

- oxyhemoglobin is diamagnetic
- deoxyhemoglobin is paramagnetic → produces field inhomogeneities
  → 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

Functional MRI (fMRI)

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 acqusition protocol:

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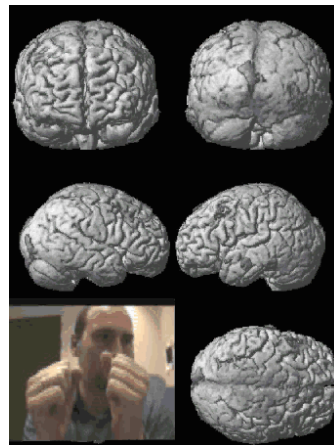- subtract these two images to visualize the active areas

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

fMRI Applications

Get insight into:

- memory
- object recognition
- language
- visual cortex
- sensorimotor cortex

Diffusion Tensor Imaging (DTI)

Capitalizes on the following facts:

- in different classes of tissue, diffusion of water molecules is different
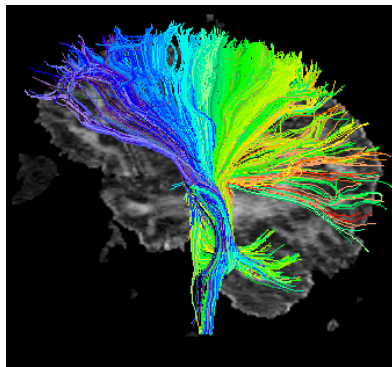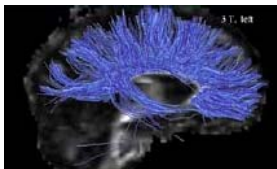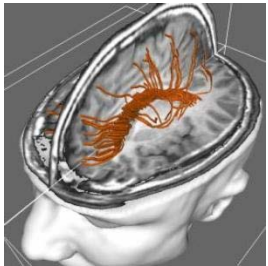- also, the diffusion is not isotropic → along fibers the diffusion is much greater than across fibers.

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
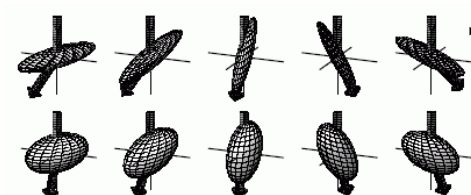
Diffusion Tensor Imaging (DTI)

Brain fibers can now be tracked using the local tensors