

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 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 $\theta_s \rightarrow 2$ knowns
- what we want is the density of $n_x \cdot n_y \cdot n_z$ individual voxels $\rightarrow n_x \cdot n_y \cdot n_z$ 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

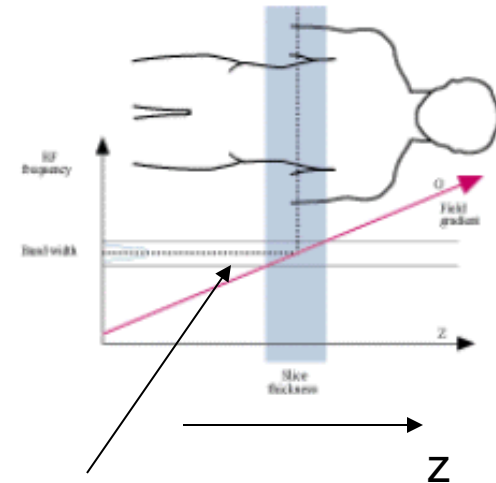
So how is this done?

- 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

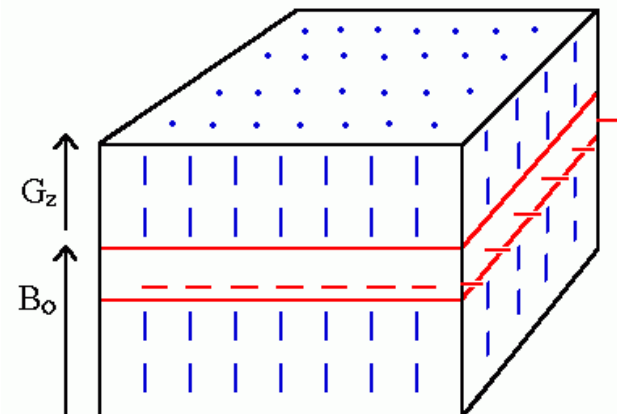
- select the slice s by coding the nuclei in that slice with a specific ω_s
- this is done by adding a magnetic field linearly rising along z (called *gradient field*)
- now only the voxels in the selected slice spin at ω_s



Step 2: RF pulse at ω_s

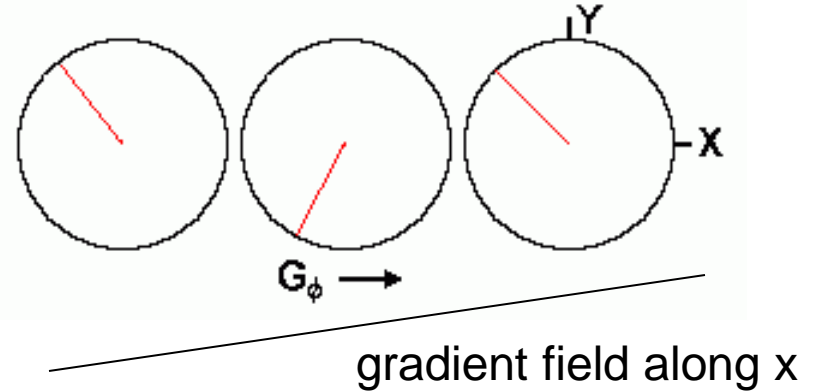
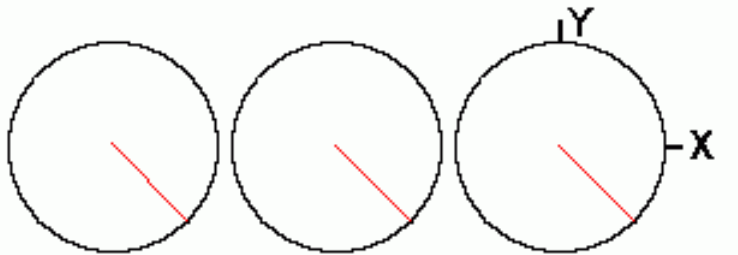
- this will only bring the nuclei in slice s into the transversal plane and sync them
- we could then measure a signal but it would still be a composite of all nuclei in the slice

Hence, more tagging is needed within the slice



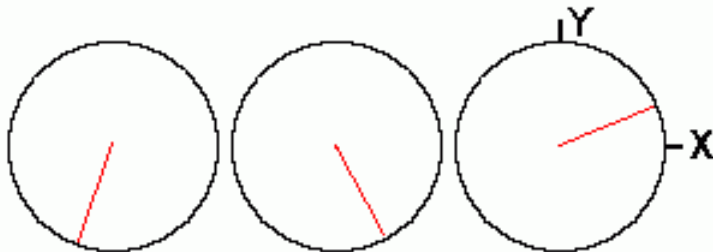
The Phase Encoding Concept

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

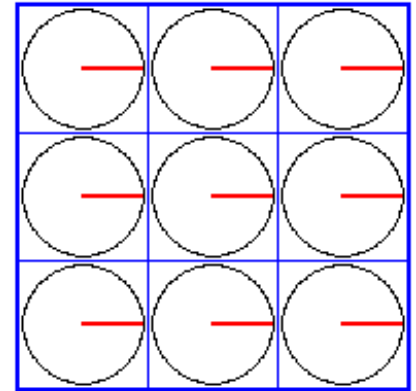
After releasing the gradient, the nuclei spin at the slice's Larmor frequency again, but an x-dependent phase shift angle remains



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

Two-Dimensional Encoding

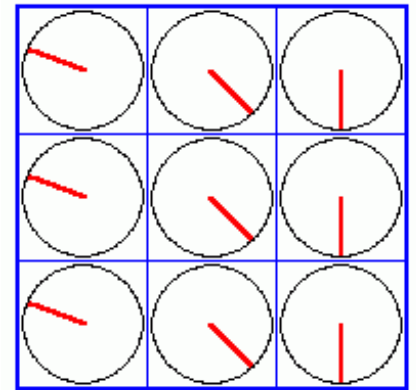
After slice selection all nuclei in the slice spin at the same rate



Apply a phase encoding (along x)

- after gradient release, the column nuclei are distinguished by phase

gradient field
along x

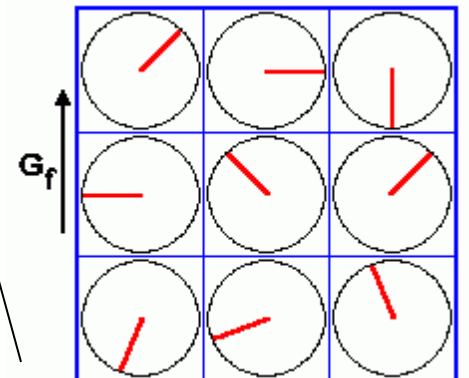


G_ϕ

Apply another gradient field (now along y)

- do not release \rightarrow frequency encoding
- now nuclei within a column can be distinguished by frequency

gradient field
along y



G_f

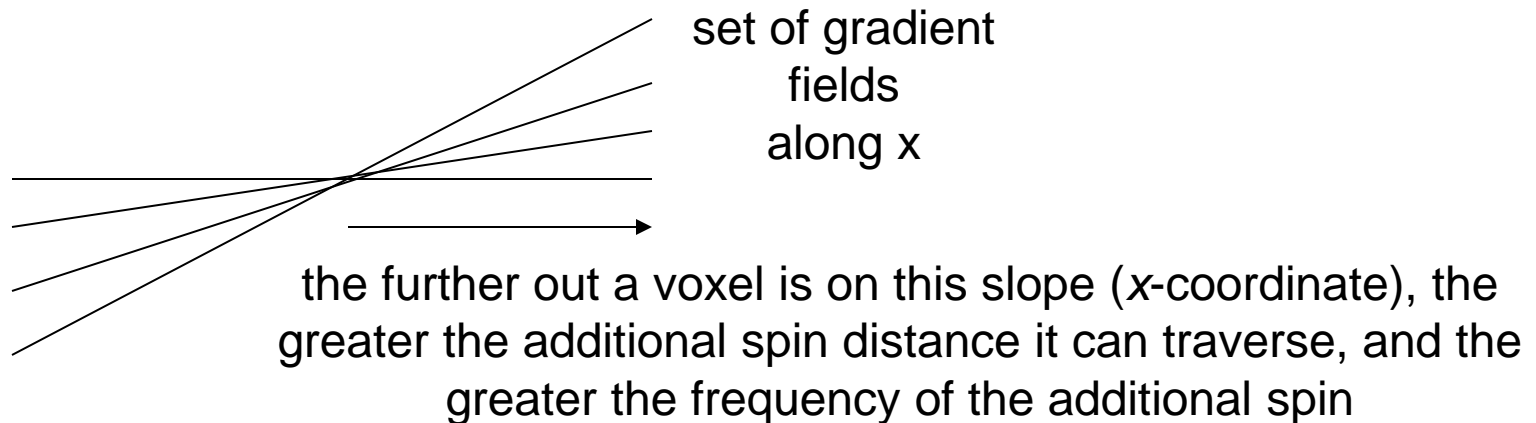
Decoding

We could measure the RF signal and perform a Fourier transform

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

Need to generate more equations

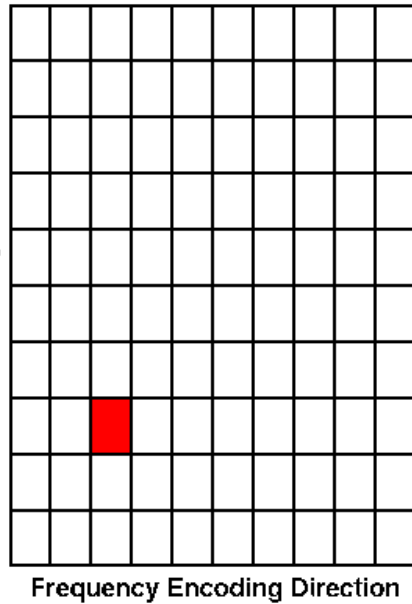
- we can get these by performing n_x different phase encodings
- for each phase encoding we use a gradient field with a different slope



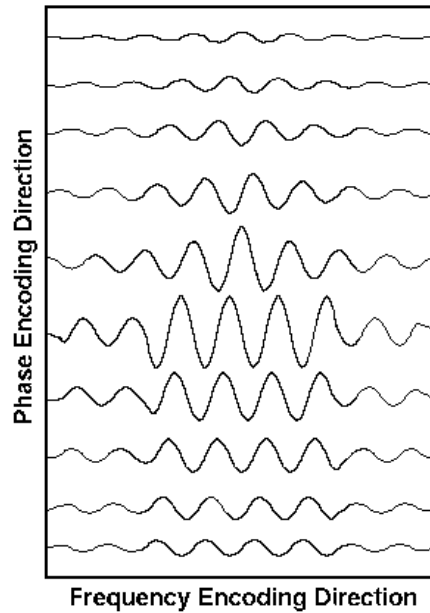
- each is then followed by a frequency encoding

Let's examine this by ways of a few examples

Voxel Decoding: Example 1



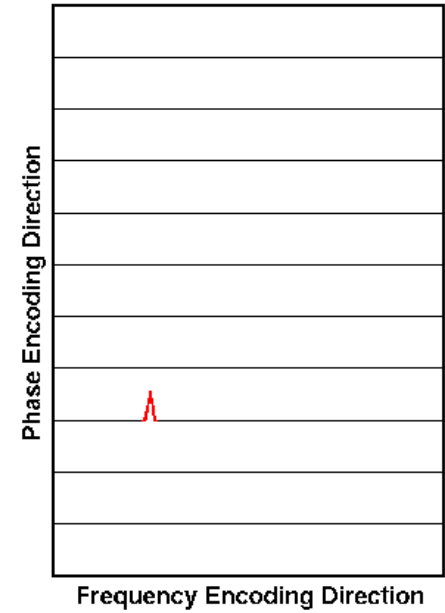
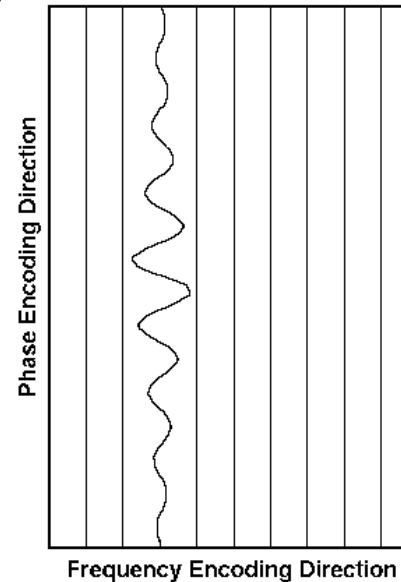
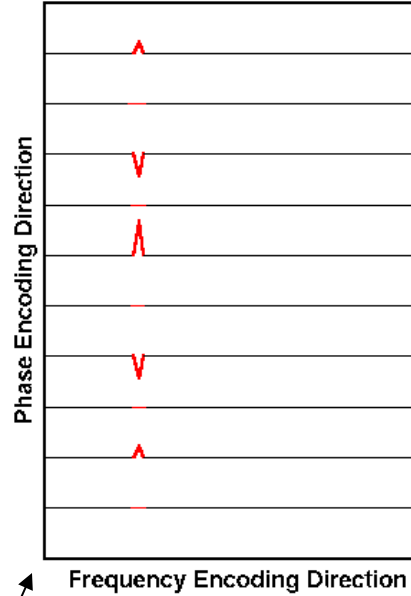
spatial domain:
only one voxel is
turned on



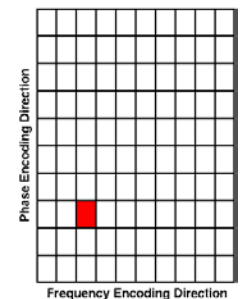
n RF measured
responses

Phase angles of the Fourier transforms
along the frequency encoding direction

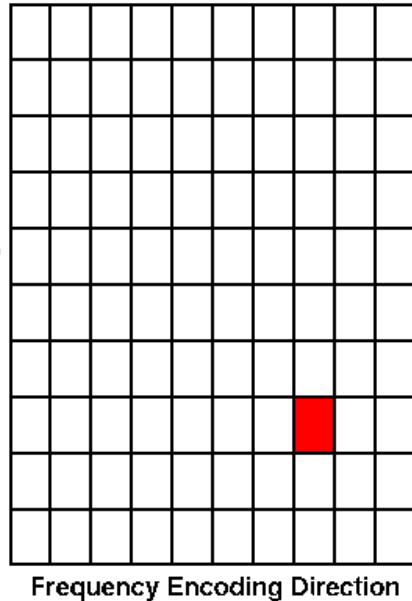
Extrapolation into continuous signal (just
for illustration)



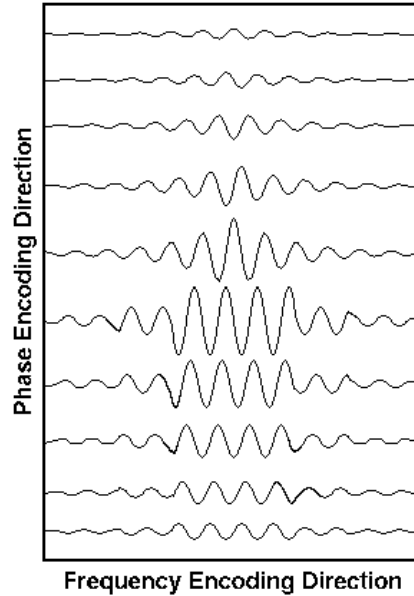
Fourier transform
along the phase
encoding direction
yields voxel
position:



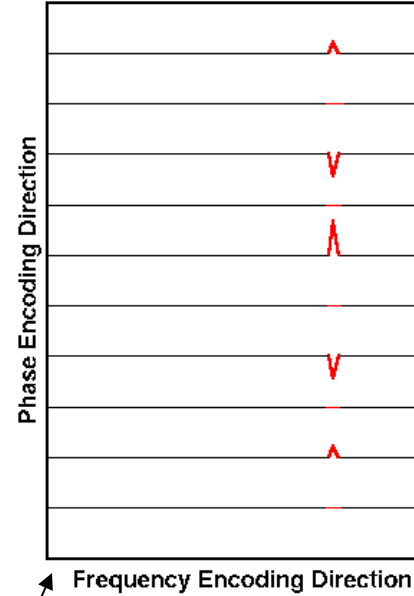
Voxel Decoding: Example 2



spatial domain:
only one voxel is
turned on

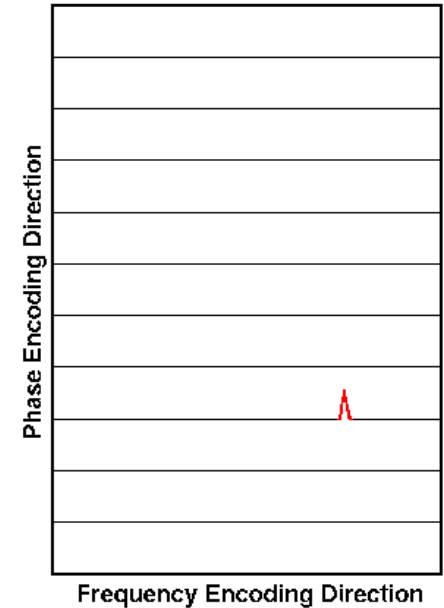
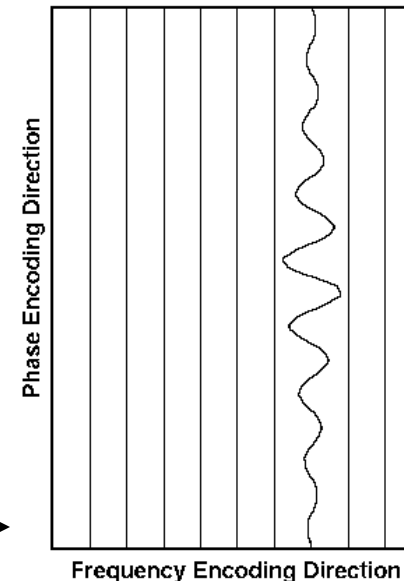


n RF measured
responses

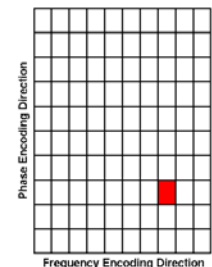


Phase angles of the Fourier transforms
along the frequency encoding direction

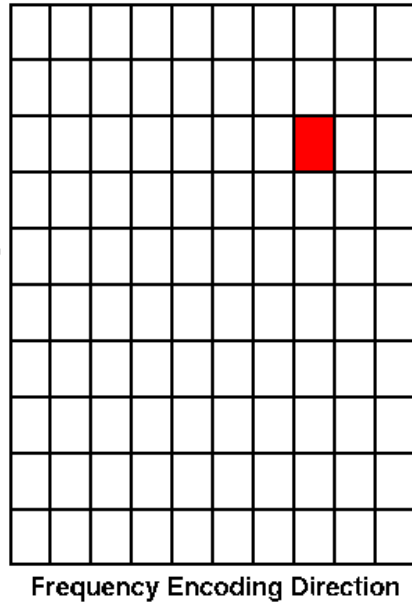
Extrapolation into continuous signal (just
for illustration)



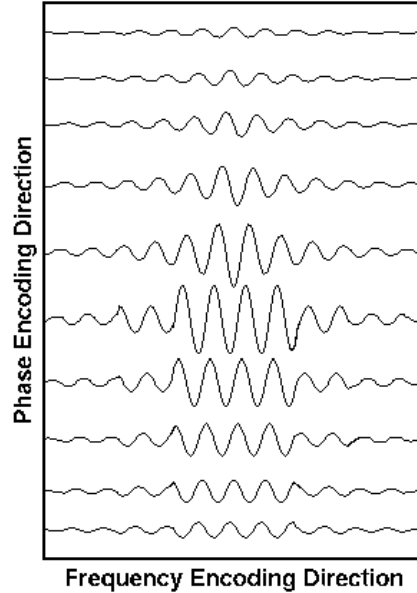
Fourier transform
along the phase
encoding direction
yields voxel
position:



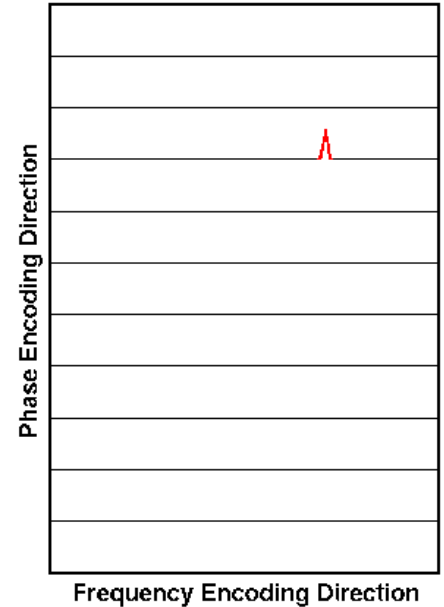
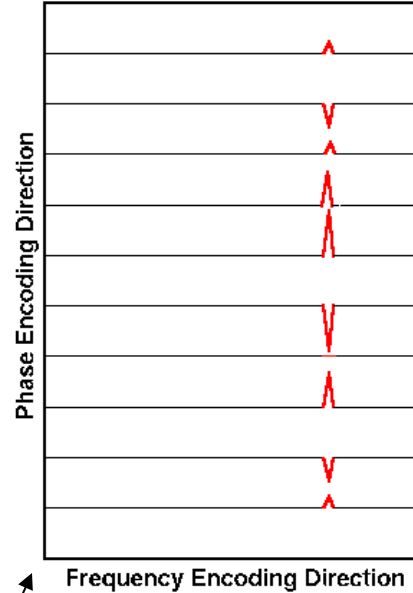
Voxel Decoding: Example 3



spatial domain:
only one voxel is
turned on



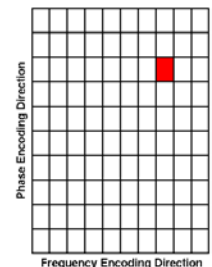
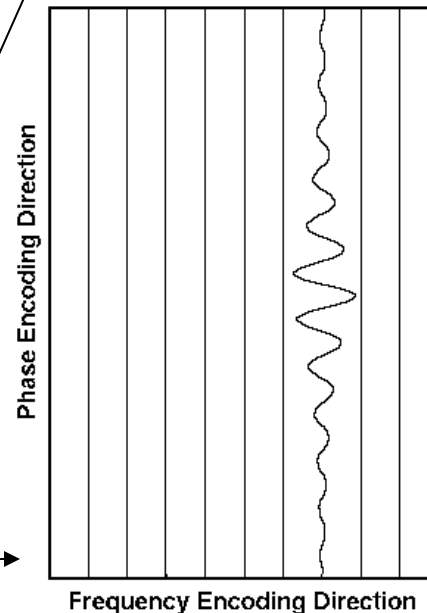
n RF measured
responses



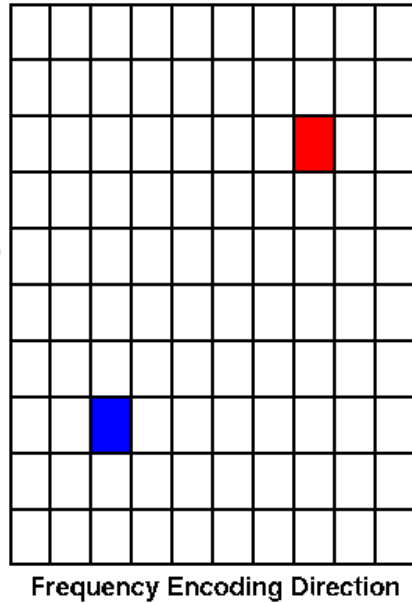
Fourier transform
along the phase
encoding direction
yields voxel
position:

Phase angles of the Fourier transforms
along the frequency encoding direction

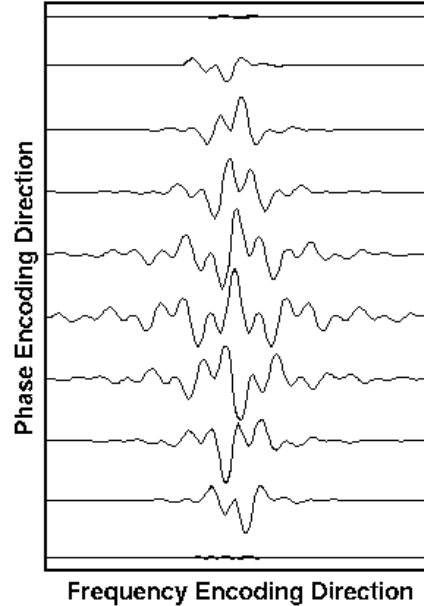
Extrapolation into continuous signal (just
for illustration)



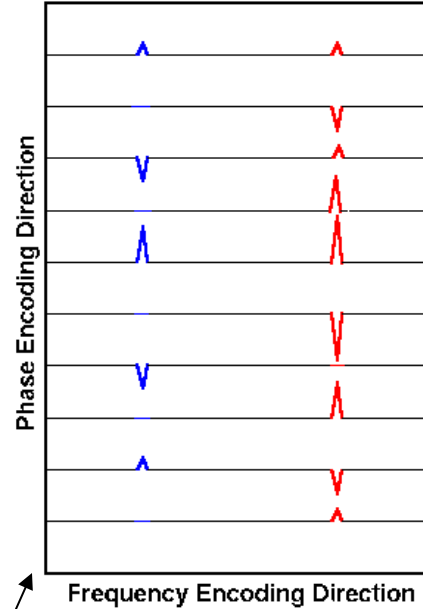
Voxel Decoding: Example 4



spatial domain:
only one voxel is
turned on

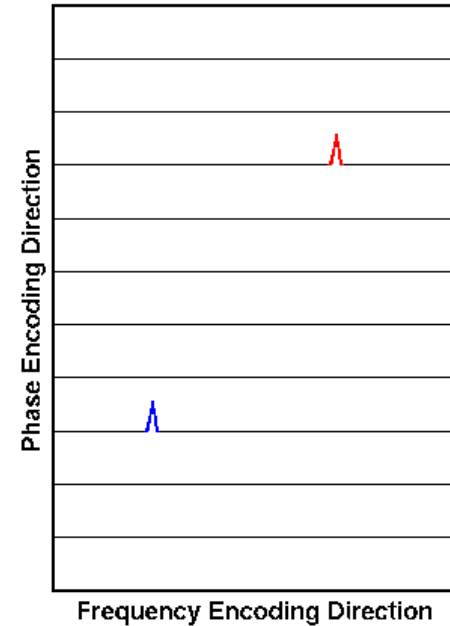
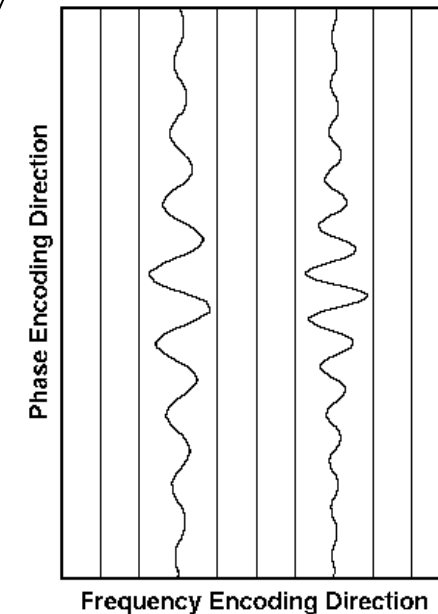


n RF measured
responses



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:

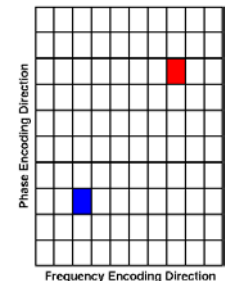
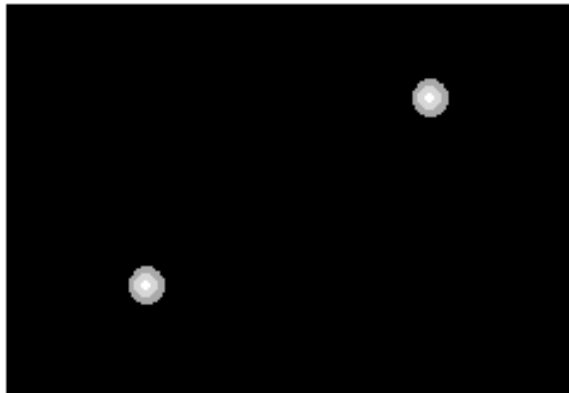
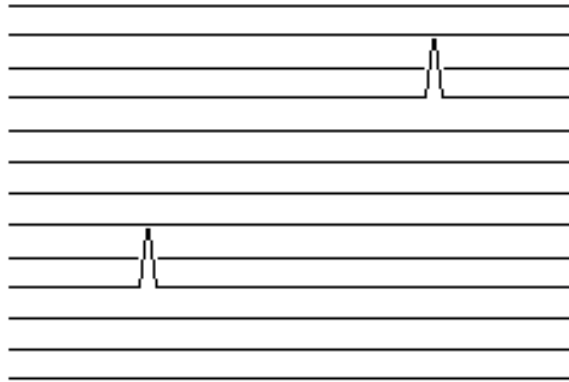


Image Generation

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



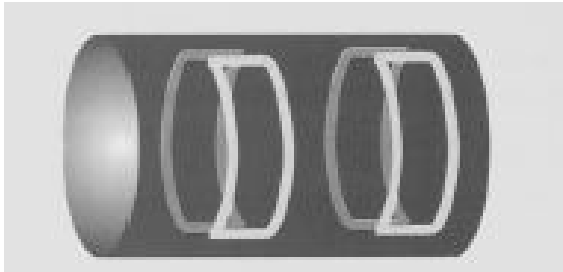
or more complex:



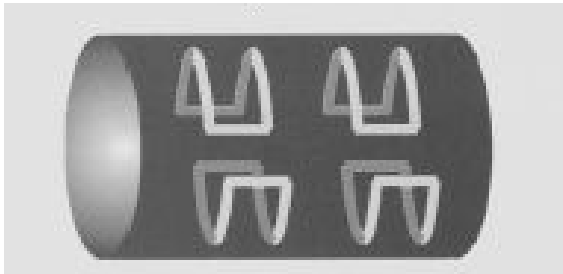
Gradient Coils

Field gradients, generated by coils, are essential in MRI

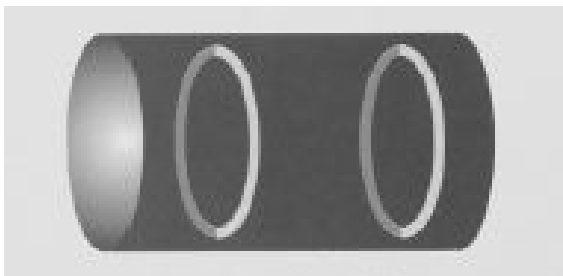
Examples:



x-direction:



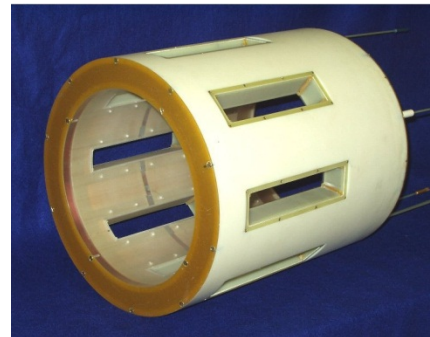
y-direction:



z-direction:



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 :

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

- the Larmor frequency of a slice is then:

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

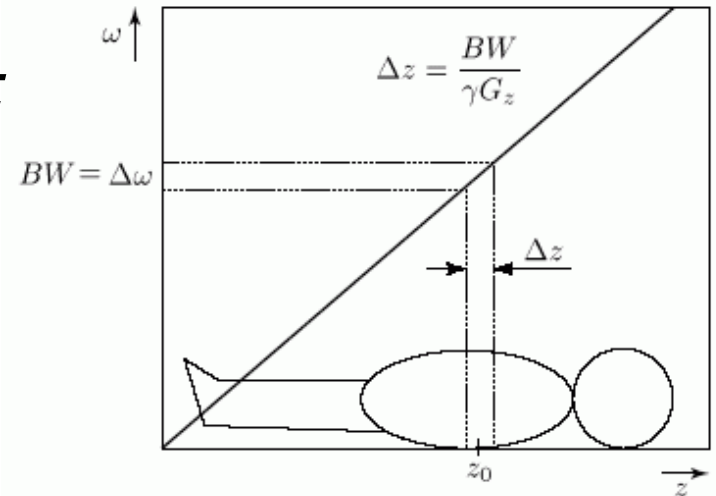
- hence, a slice of thickness Δz will resonate at frequencies:

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

- 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: $\Delta z = 2 \text{ mm}$ ($B_0=1.5\text{T}$), $\Delta z = 3 \text{ mm}$ ($B_0=1\text{T}$)



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

Slice Selection: Example

Dynamic cardiac sequence

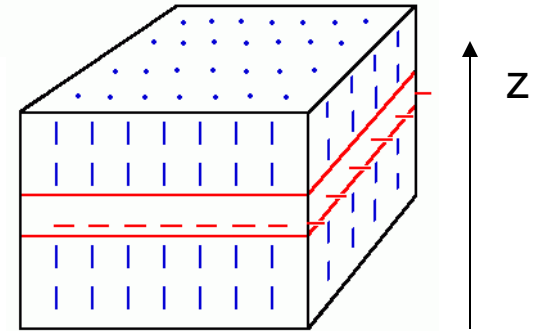
- four cardiac chambers together with the heart valves in a plane parallel to the cardiac axis



Position Encoding: Spin Differentiation

The slice selection RF pulse puts all nuclei within Δz 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 ω_0 , in a synchronized fashion



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

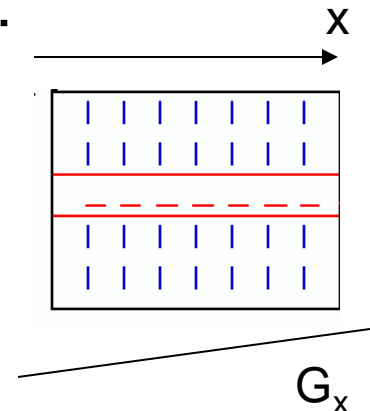
For encoding, we apply a field gradient G_x along x:

- this changes the transverse component frequency to:

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

- in complex notation:

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



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

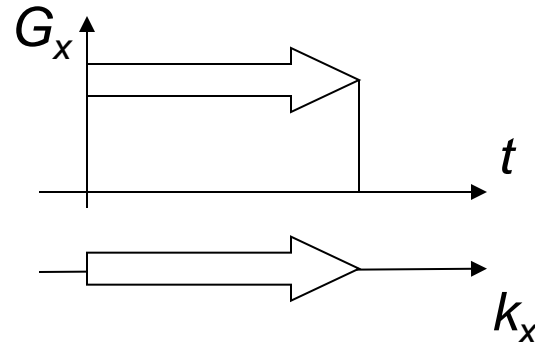
Recall 1D Fourier Transform: $F(k_x) = \int_{x=-\infty}^{+\infty} f(x) e^{-i2\pi k_x x} dx$

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

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)

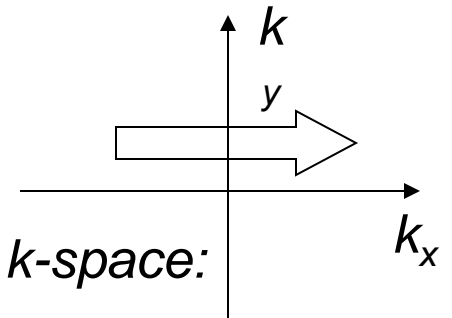
A generalization of what we have just seen

- 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:

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

- 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 $h(t, T_2)$:

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

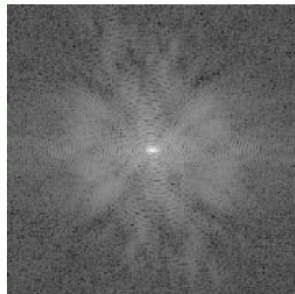
MRI seeks techniques where the returned values at (x, y, z) all have *the same* relaxation effects

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

The K-Theorem (3)

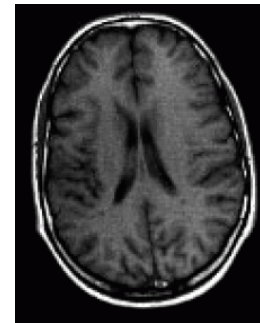
We have seen that the time signal $s(t)$ is equivalent to the Fourier Transform (FT) of the desired image

- as mentioned, this data collection space is called *k-space*
- an inverse FT will produce the desired image



k-space image

2D Fourier
transform



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

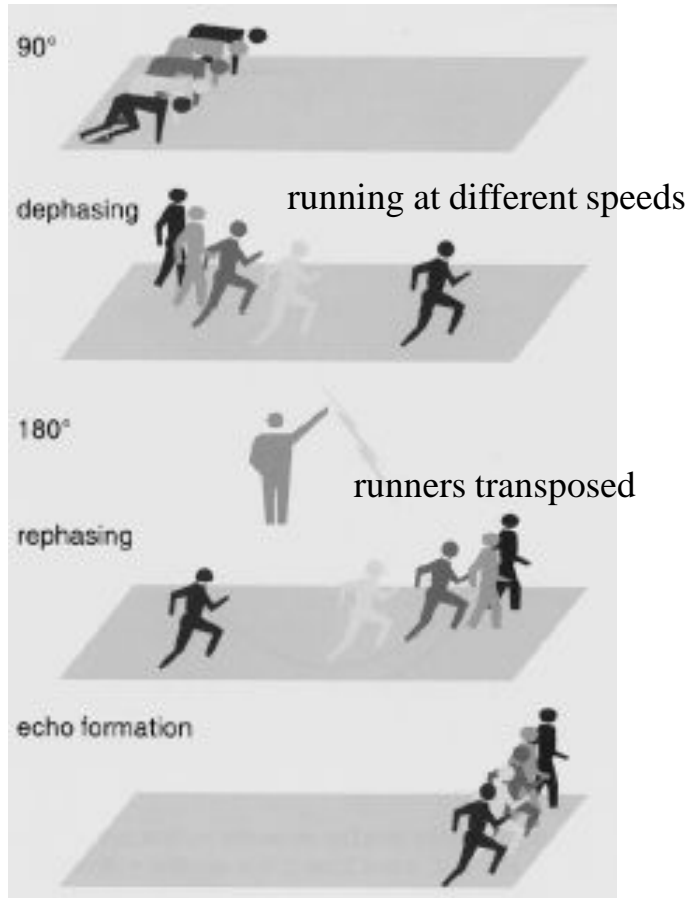
3. Magnetic field gradients:

- reversible, time constant $T_2^{**} < T_2^*$
- undone with reversed gradient fields

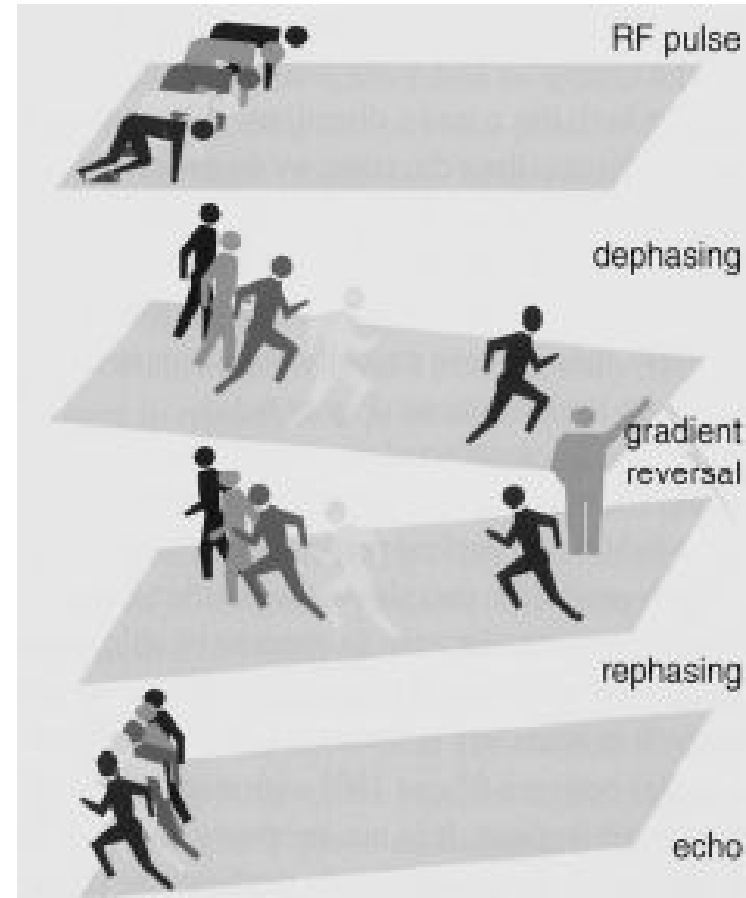
Rephasing: Runners Equivalent

Rephasing is the process of undoing the effects of dephasing

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

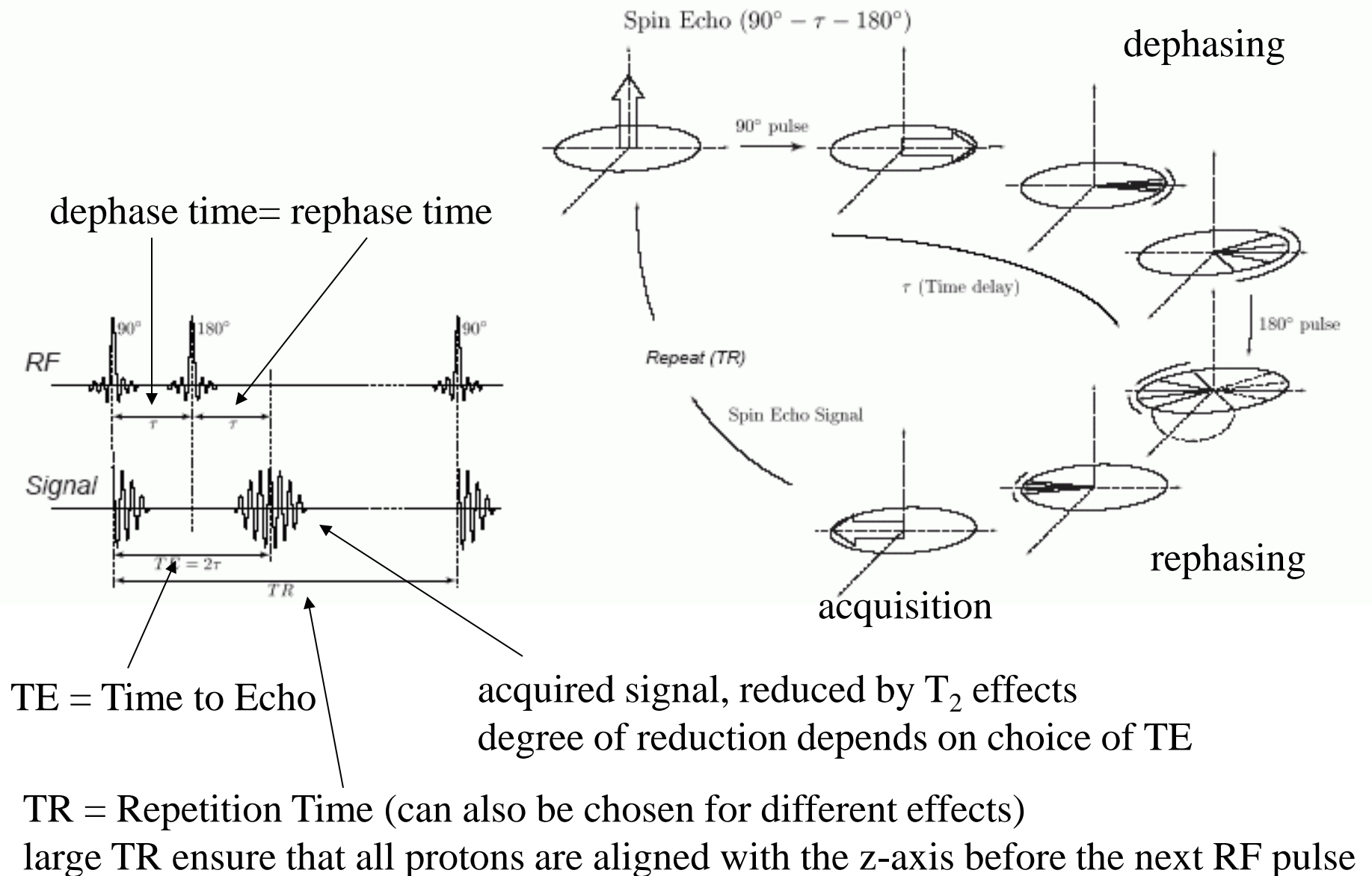


180° RF pulse



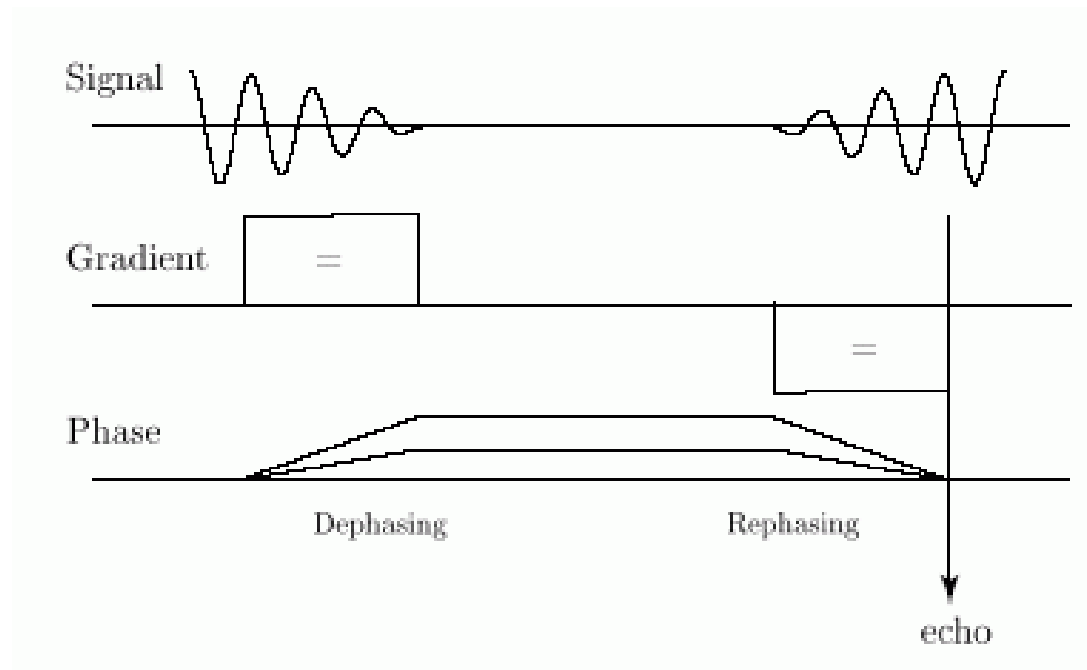
Gradient reversal

Rephasing: 180 RF Pulse



Rephasing: Gradient Reversal

Opposite field gradients of same length and strength



The Art of Choosing TR and TE

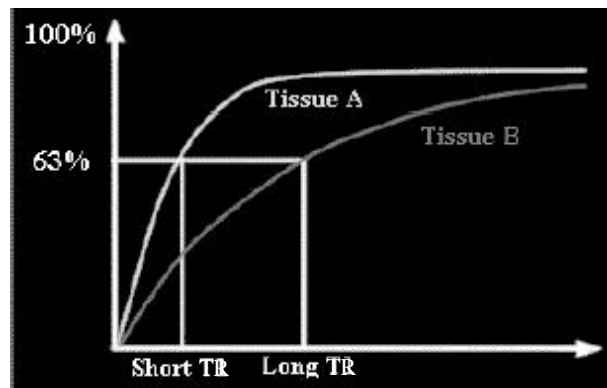
TR and TE can be varied to bring out proton density (PD), T1, and T2 properties of the tissue

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 $t=0$

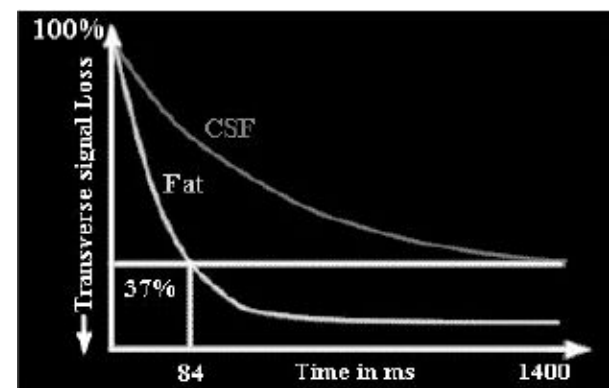
MRI seeks to bring out tissue contrast as manifested by differences in PD, T1, or T2 properties (T1/TR, T2/TE ratios)

- 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



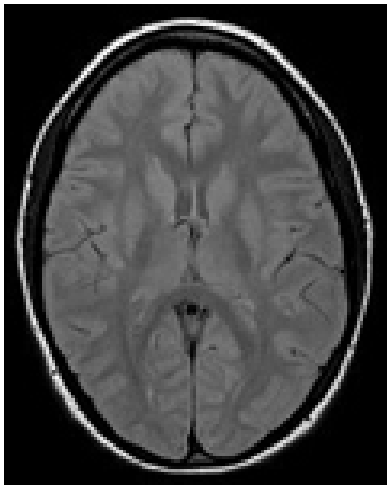
Examples

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



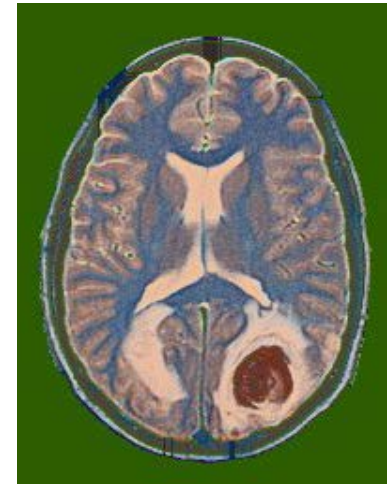
proton density



T1-weighted



T2-weighted



T1/T2 fused

T2 Weighting

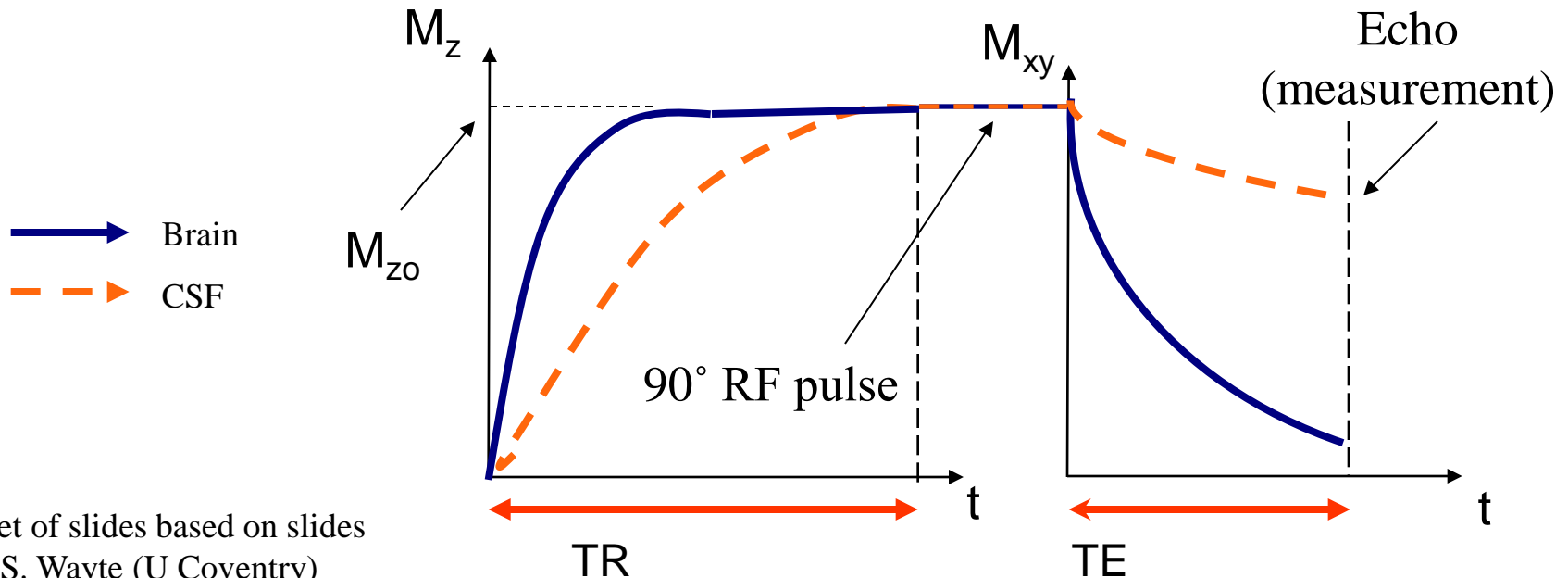
Long TR

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

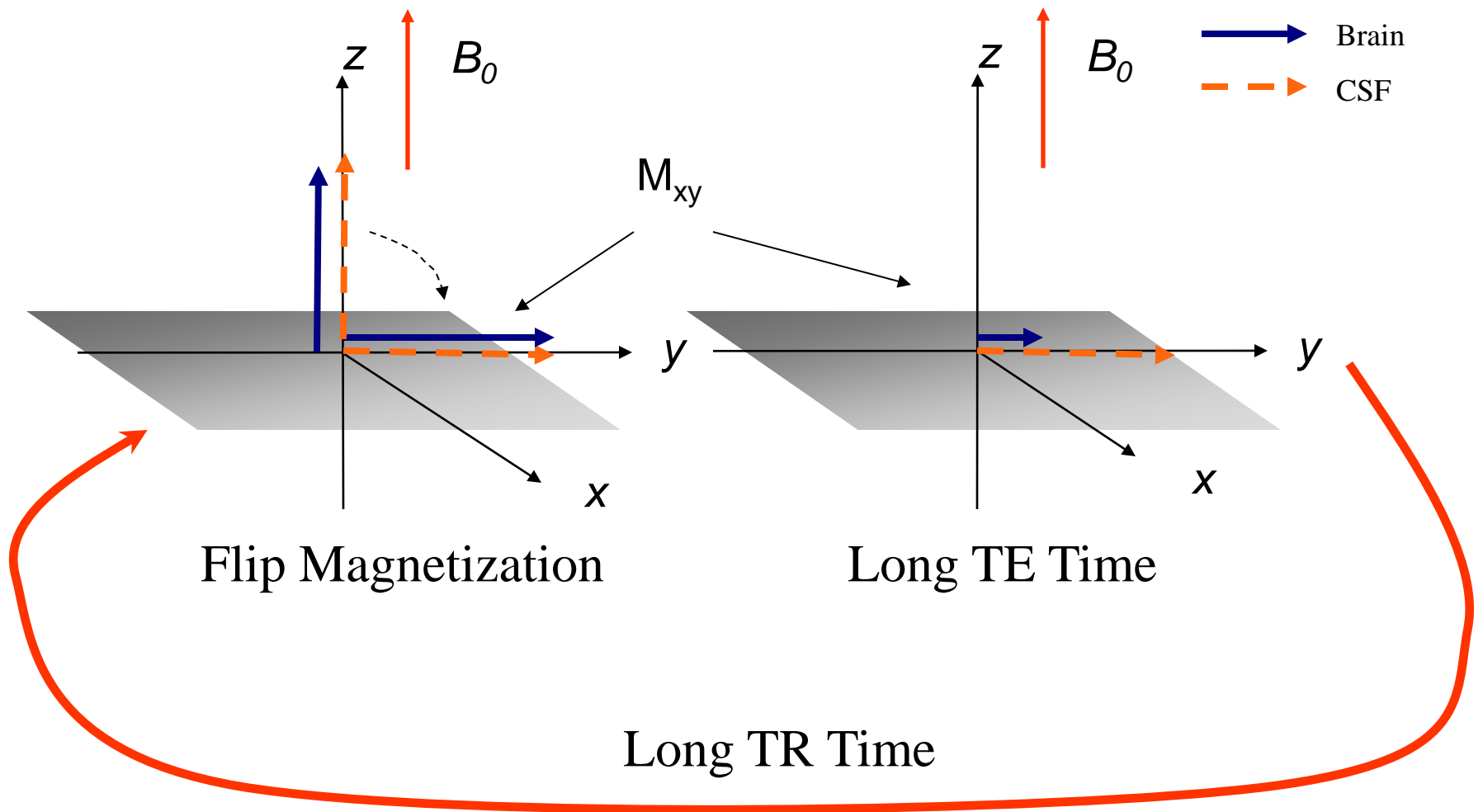
Long TE

- to give T2 weighting time to develop

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



T2 Weighting

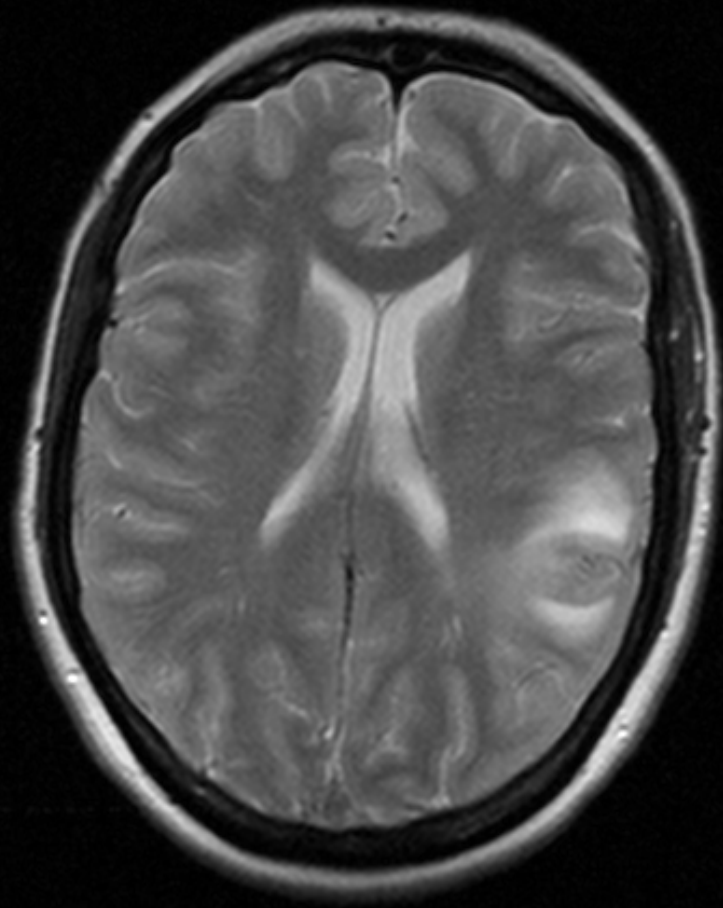


T2 Weighting: Examples

Anonymous, F, 1966.08.07
L80702

Univ Hosp Cov & Wark

TR = 3070ms TE = 92ms



2002.07.01
Acq: 1, Image: 26

Anonymous, M, 1954.12.07
G05268

Walsgrave Hospital
MR mrc

TR = 4000ms

TE = 132ms

2000.01.18

Acq: 1, Image: 23



T1 Weighting

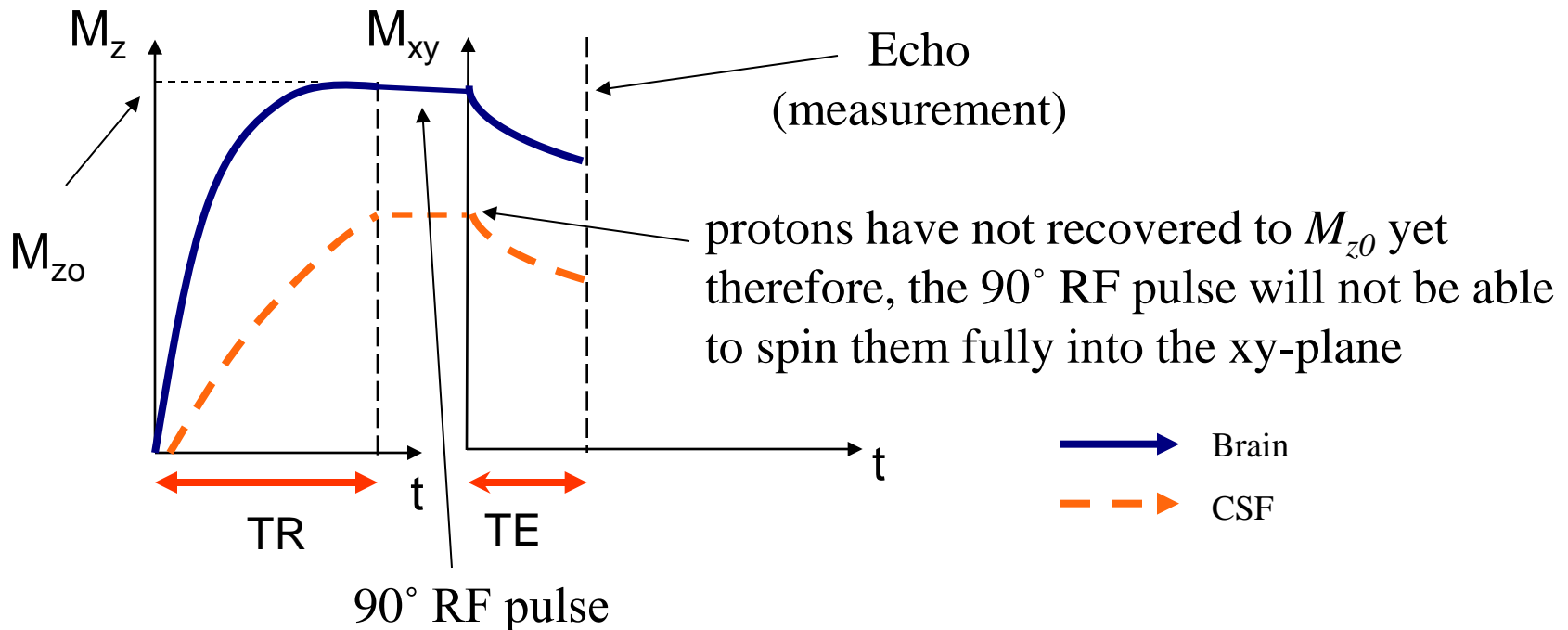
Short TR

- to bring out differences in T1 between tissues

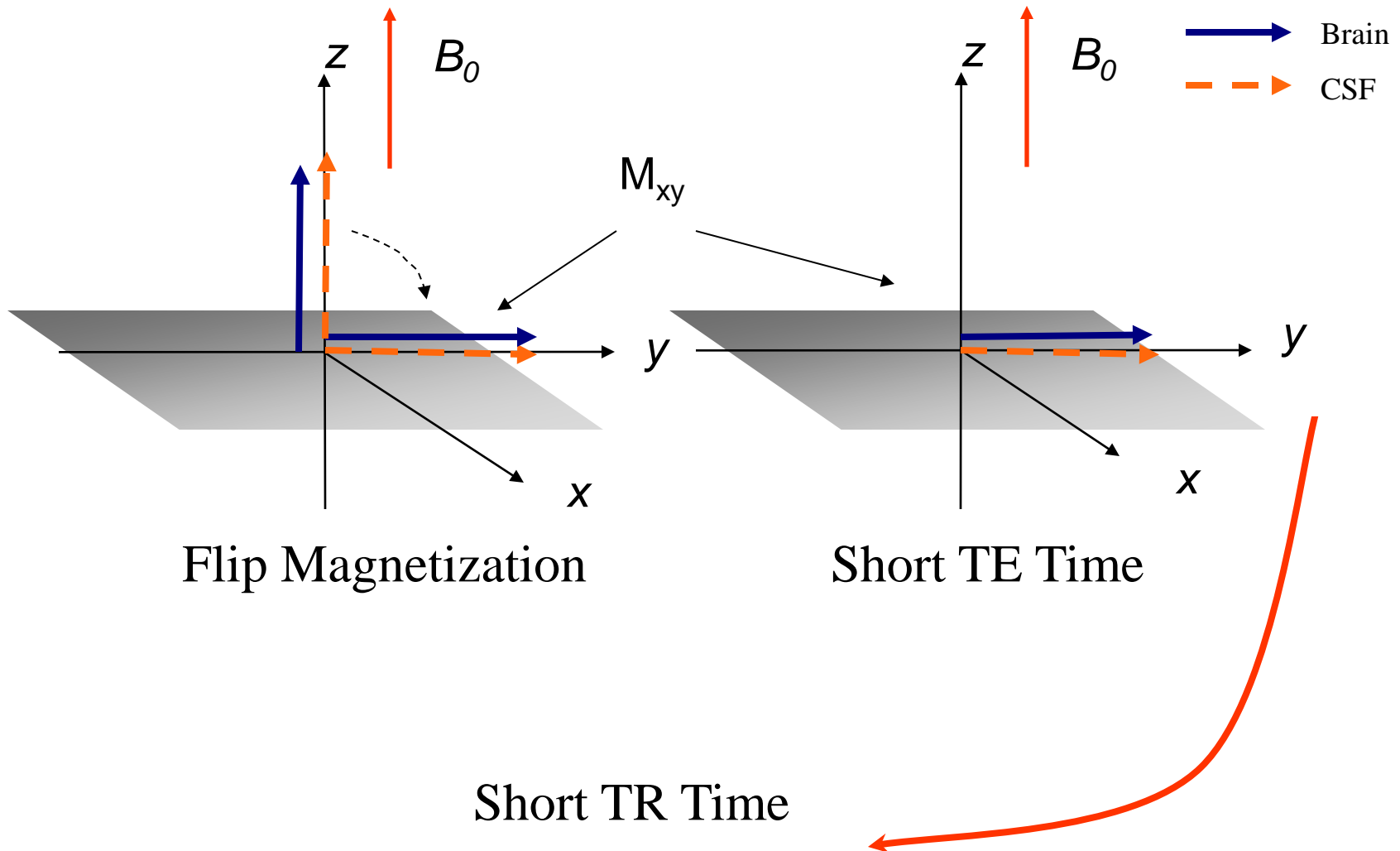
Short TE

- so differences in T2 do not have time to appear

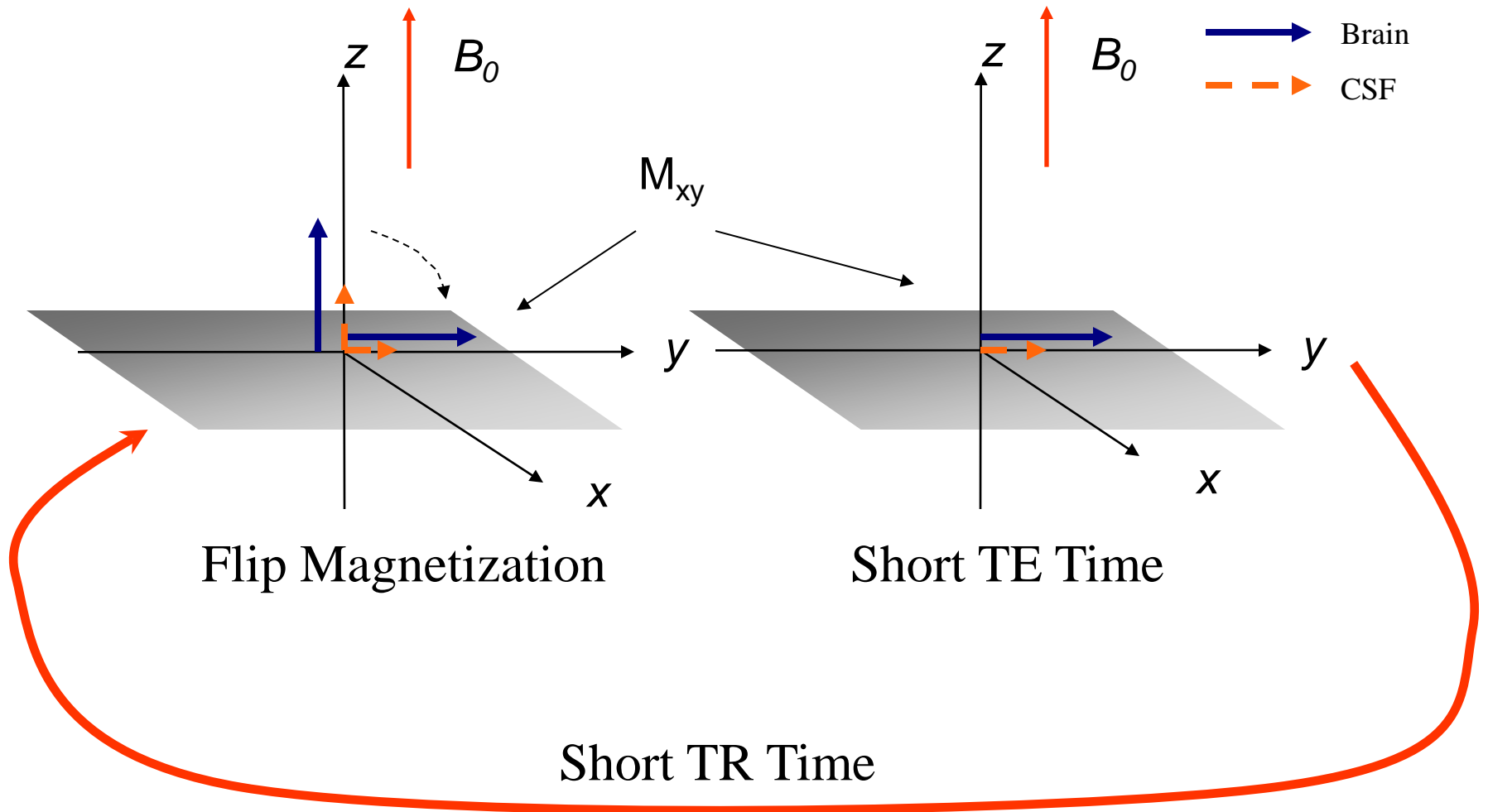
$$M_{xy}^{T1} = M_{xy0} (1 - e^{-\frac{TR}{T1}}) e^{-\frac{TE}{T2}} \Rightarrow M_{z0} (1 - e^{-\frac{TR}{T1}}) \Rightarrow M_{z0} \frac{TR}{T1}$$



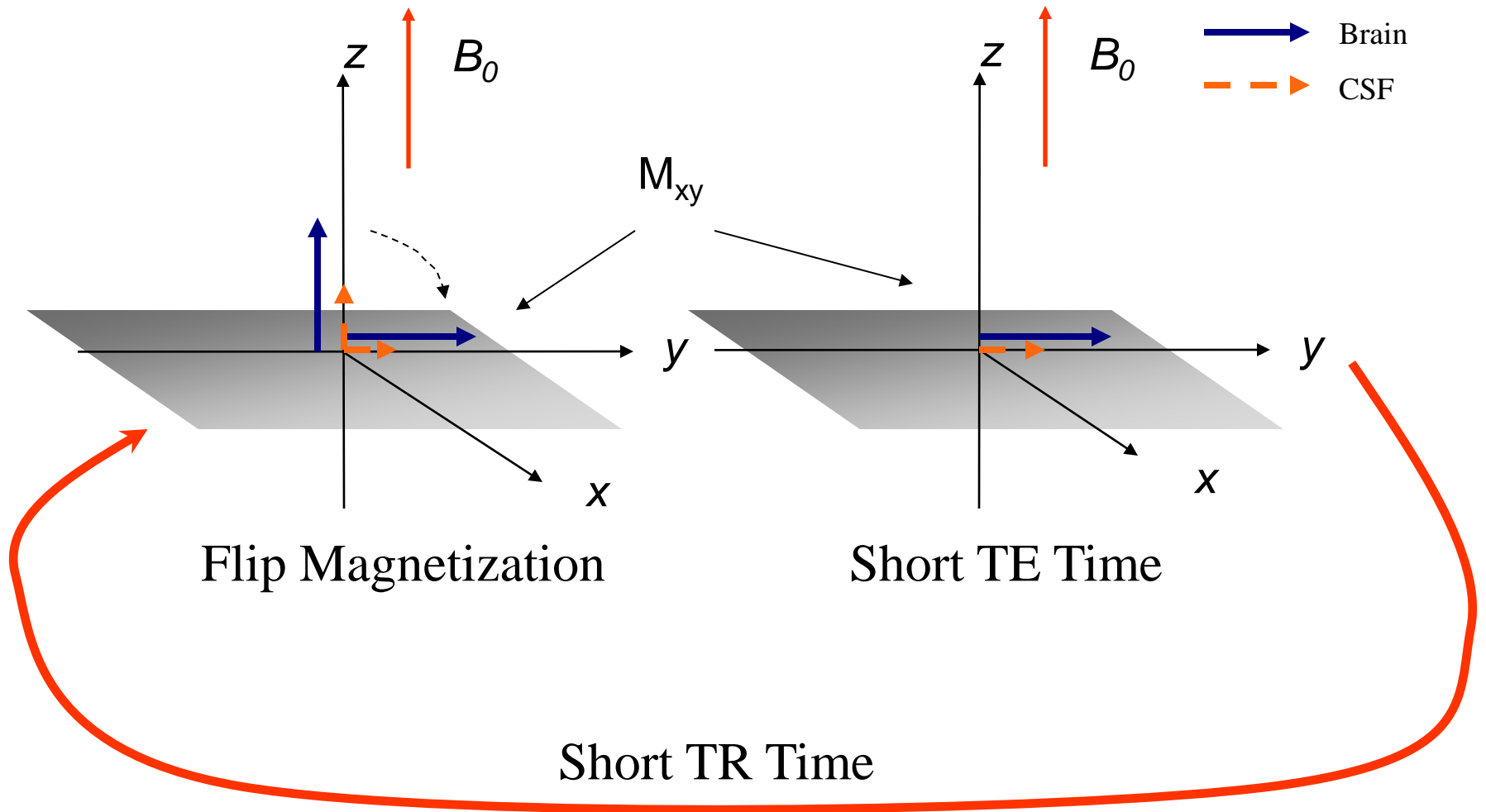
T1 Weighting: First Time Around



T1 Weighting: Second Time Around



T1 Weighting: All Subsequent Rounds



T1 Weighting: Examples

TR=525ms TE=15ms

Anonymous, F, 1958.11.07
C68470

Walsgrave Hospital
MR mrc



2000.06.12
Acq: 1, Image: 74

Anonymous, M, 1954.12.07
G05288

Walsgrave Hospital
MR mrc

TR = 562ms

TE = 20ms



2000.01.18

DJB Acq: 1, Image: 10

T1 vs. T2 Weighting (1)

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



T1 vs. T2 Weighting (2)

Anonymous, M, 1954.12.07
G05268

TR = 562ms

TE = 20ms



Walsgrave

Anonymous, M, 1954.12.07
G05268

TR = 4000ms

TE = 132ms



Walsgrave Hospital
MR mrc

2000.01.18
Acq: 1, Image: 10

2000.01.18
Acq: 1, Image: 23

T1 vs. T2 Weighting (3)

TR=525ms TE=15ms

Anonymous, F, 1956.11.07
C66470

Walsgrave Hospital
MR mrc



2000.06.12
Acq: 1, Image: 74

DJB

TR=2500ms TE=85ms

Anonymous, F, 1956.11.07
C66470

Walsgrave Hospital
MR mrc



2000.06.12
Acq: 1, Image: 53

DJB

Proton Density Weighting

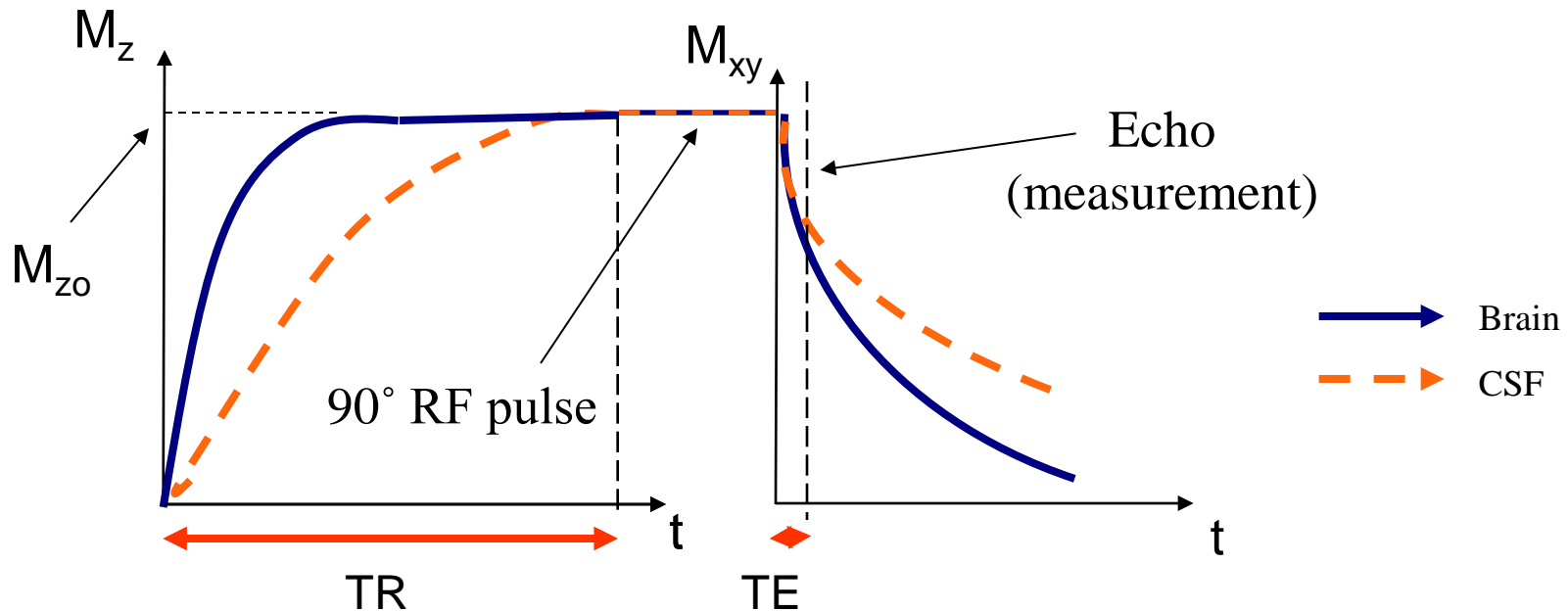
Long TR

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

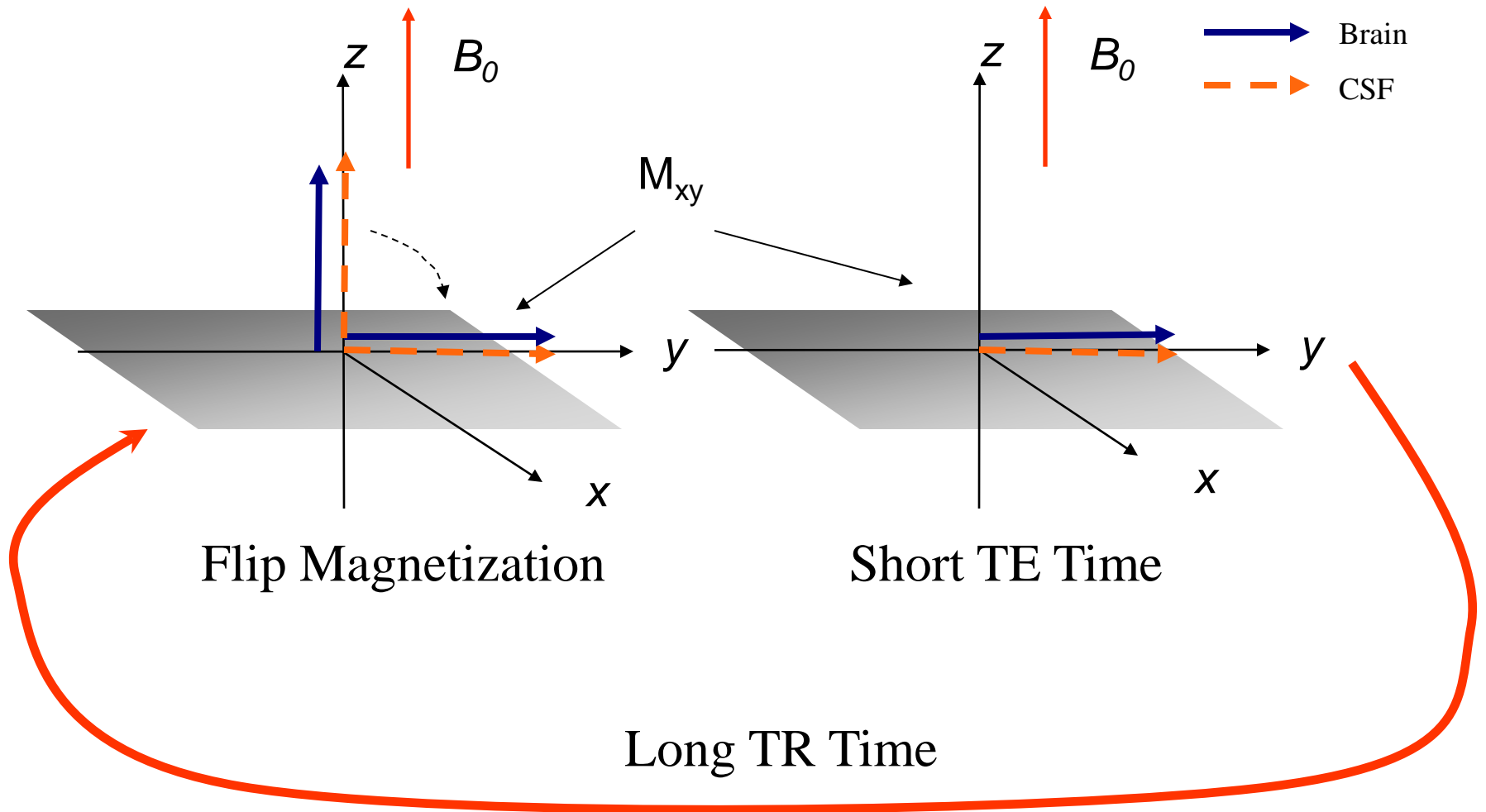
Short TE

- 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



Proton Density vs. T2 (1)

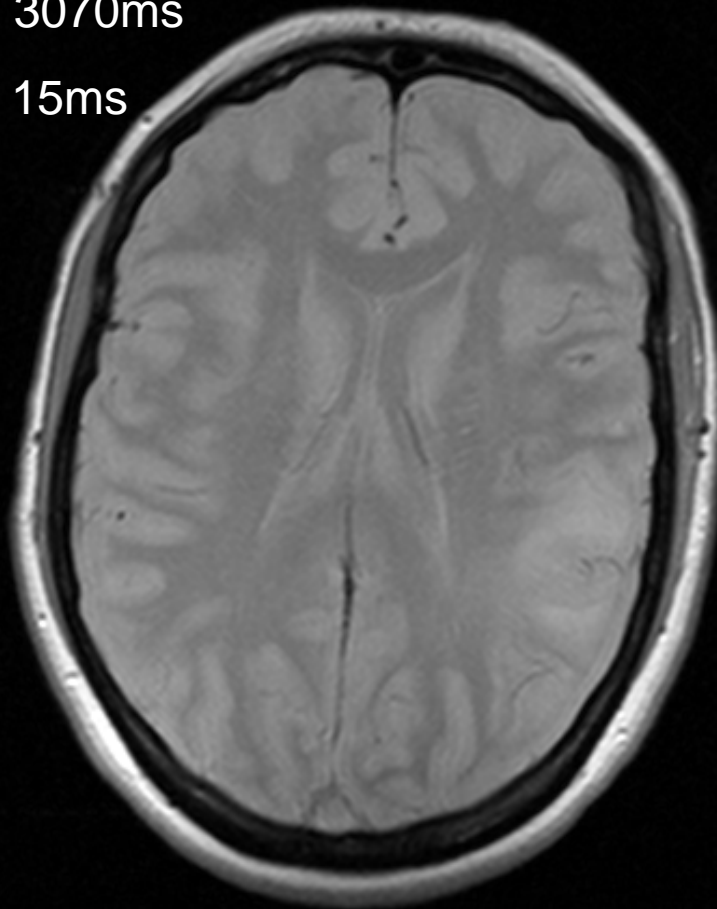
Anonymous, F, 1966.08.07
L80702

Univ Hosp Cov & Warks NHS1 Anonymous, F, 1966.08.07
MR L80702

Univ Hosp Cov & Warks NHST
MR

TR = 3070ms

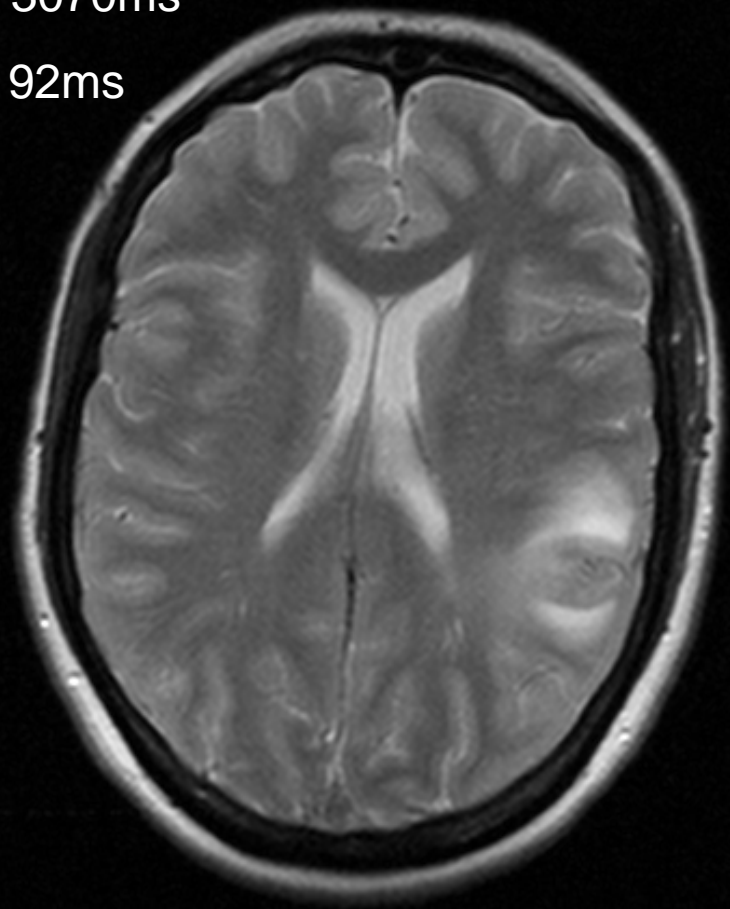
TE = 15ms



2002.07.01
Acq: 1, Image: 7

TR = 3070ms

TE = 92ms



2002.07.01
Acq: 1, Image: 26

Proton Density vs. T2 (2)

Anonymous, F, 1966.08.07
L80702

Univ Hosp Cov & Warks NHST
MR L80702

Univ Hosp Cov & Warks NHST
MR

TR = 3070ms

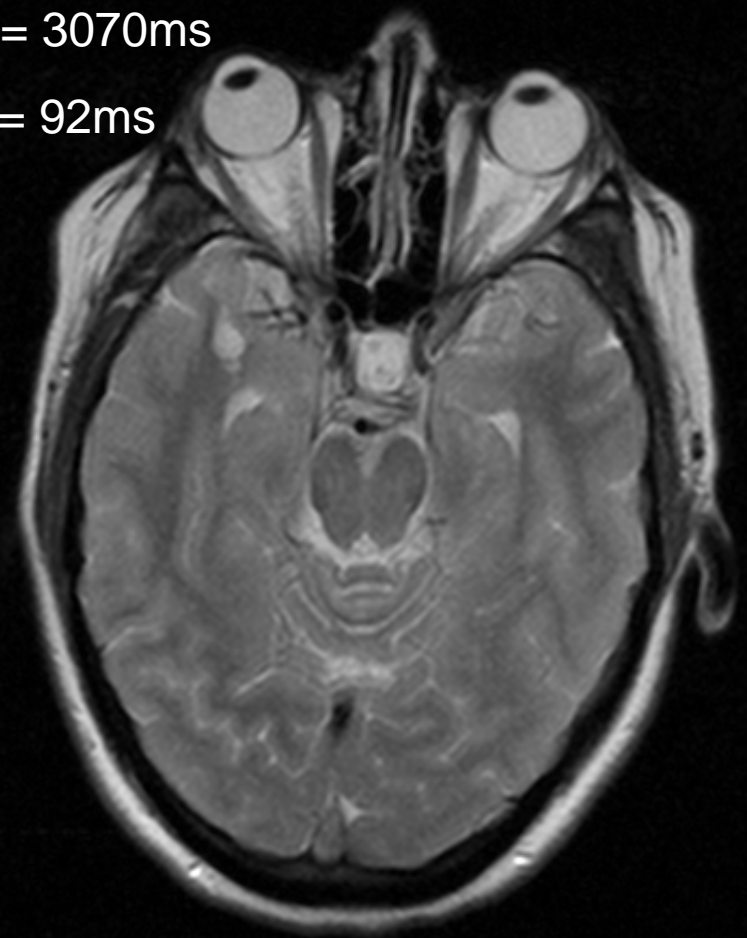
TE = 15ms



2002.07.01
Acq: 1, Image: 12

TR = 3070ms

TE = 92ms

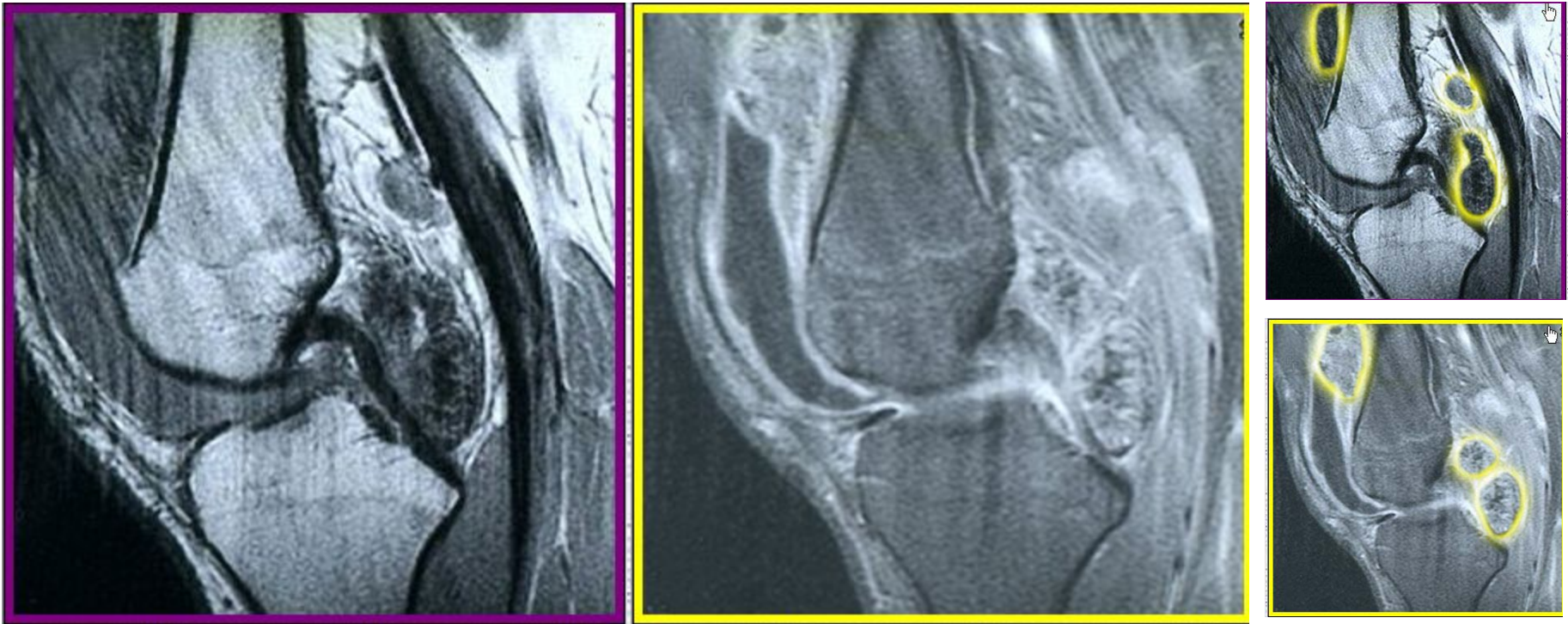


2002.07.01
Acq: 1, Image: 31

Proton Density vs. T1

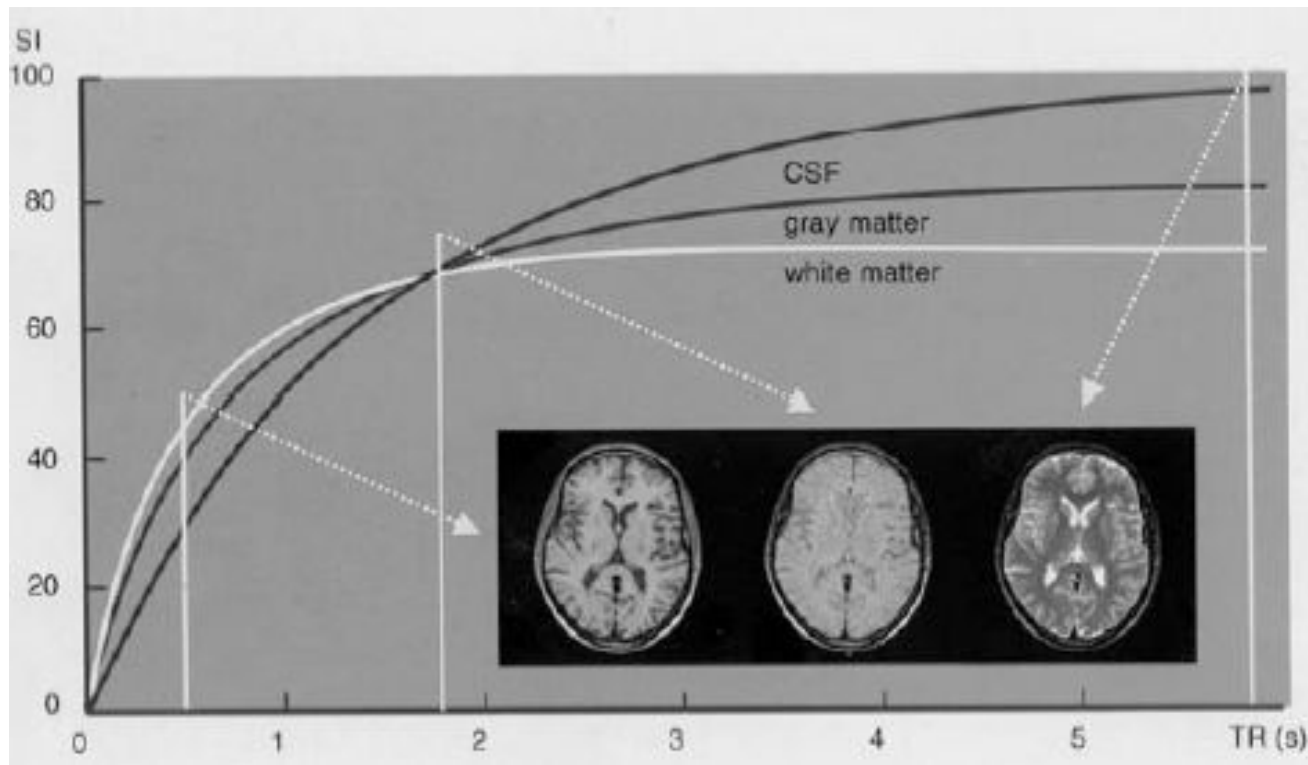
Knee with Pigmented Villonodular Synovitis (PVNS)

- example shows that PD and T1 weighting creates contrast for complementary features



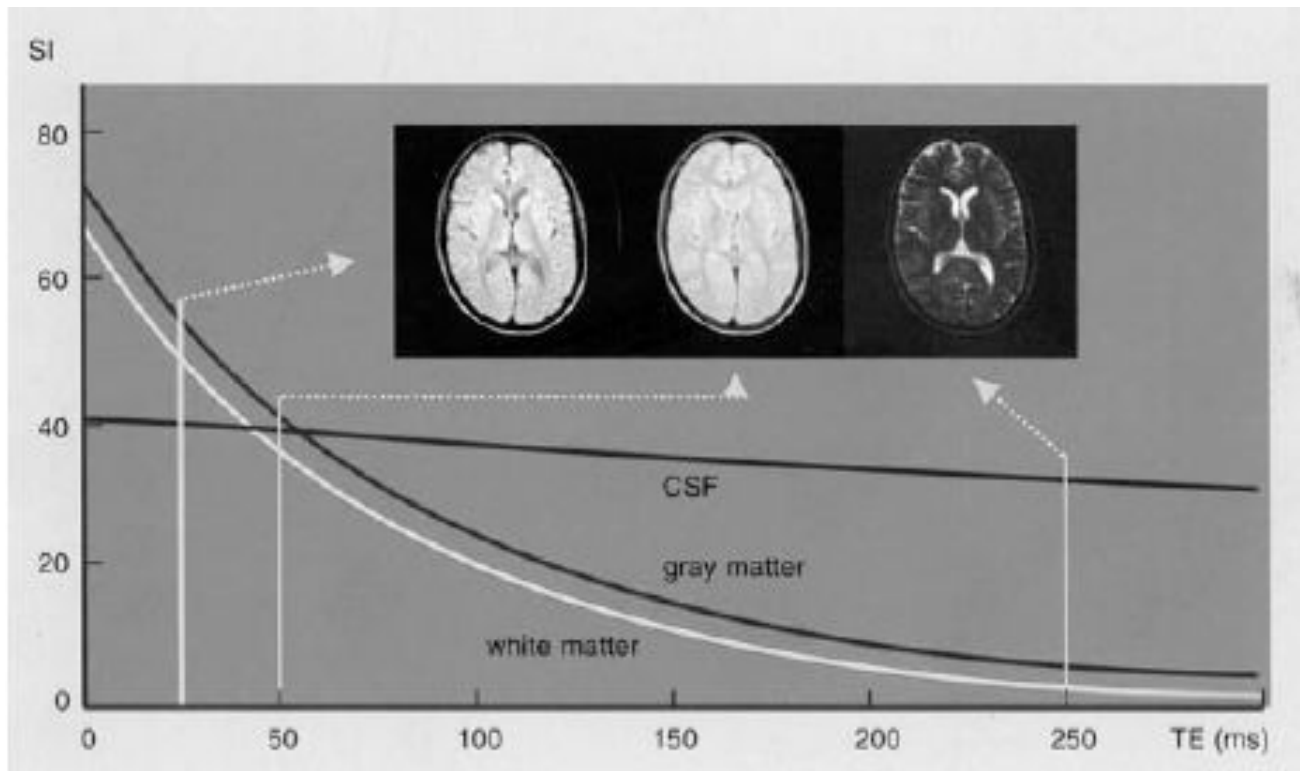
Contrast Dependence on TR

Use TR to maximize contrast among tissues of interest



Contrast Dependence on TE

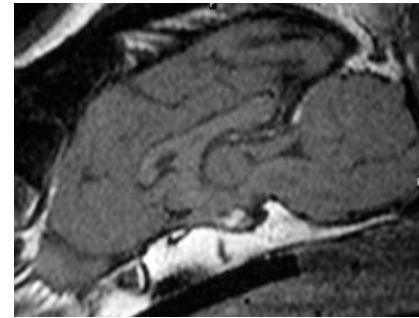
Use TE to maximize contrast among tissues of interest



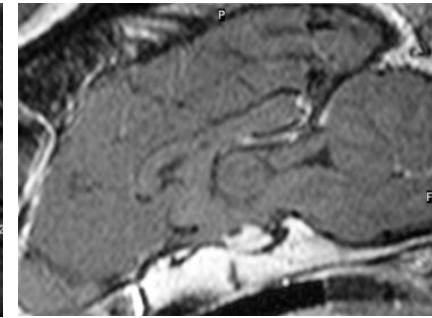
Other Techniques

Gadolinium injection

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



without



with Gadolinium

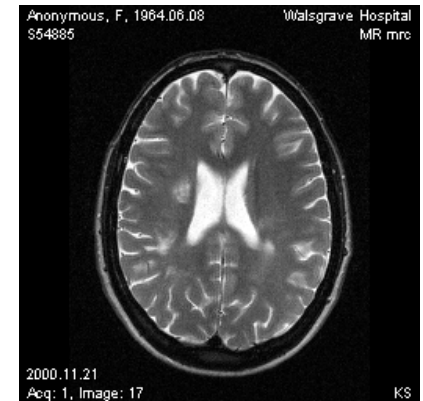
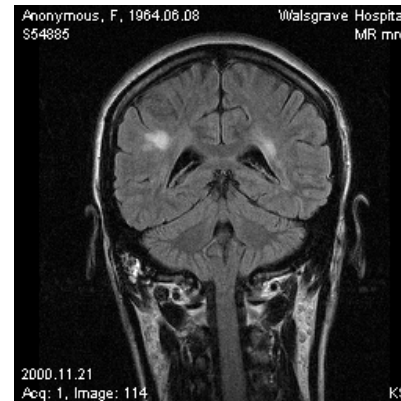
Inversion recovery

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

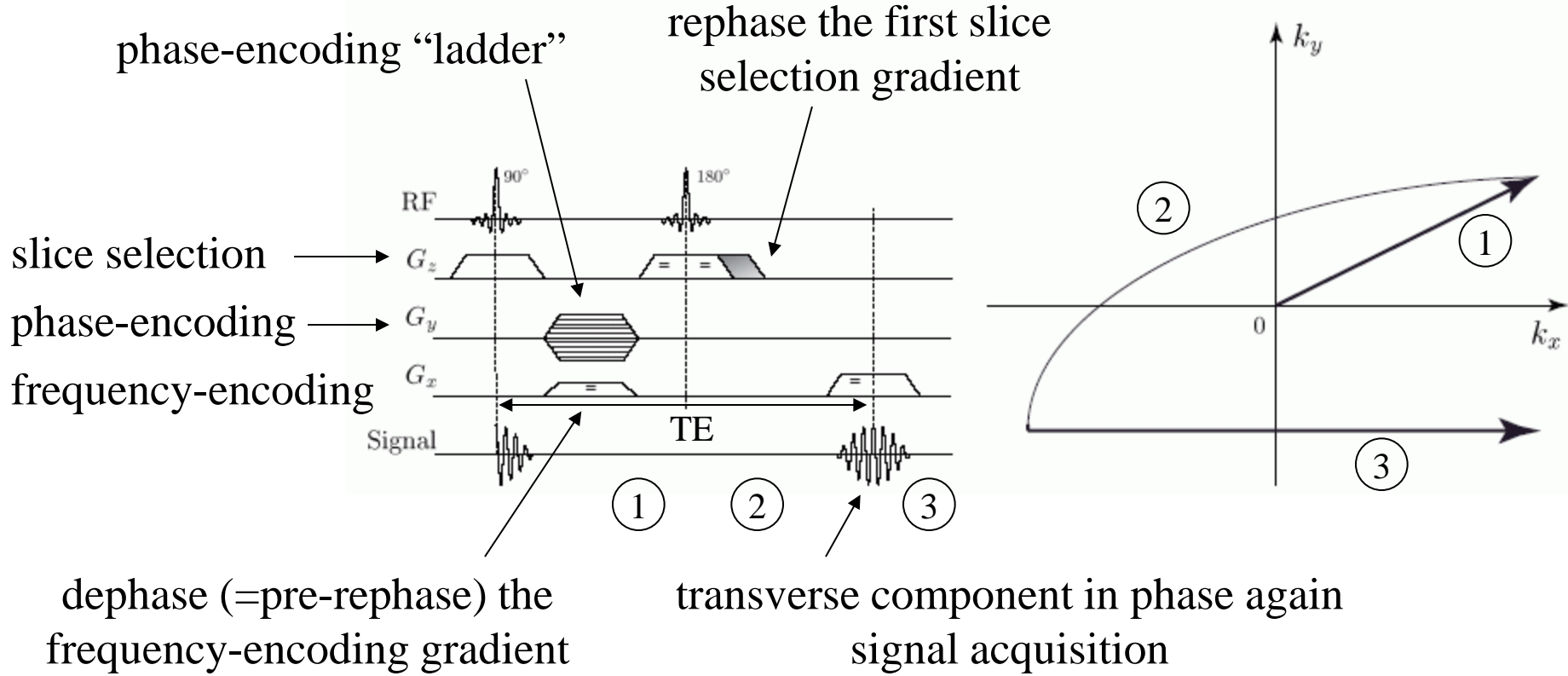
FLAIR=Fluid Attenuated Inversion Recovery

- used often to enhance lesions inside the brain in areas close to the fluid surrounding it

And many more...



Sequences: Spin-Echo

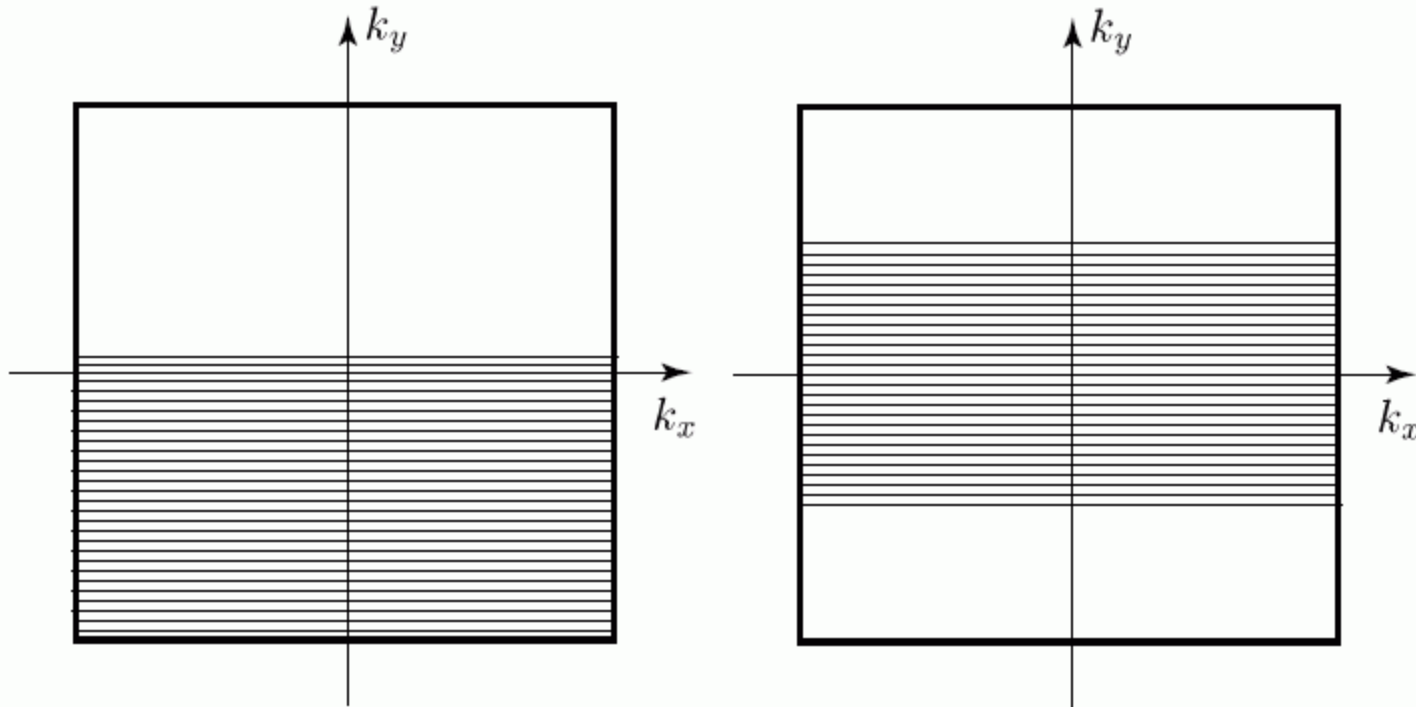


Phase shift: $\phi(y) = \gamma G_y y T_{\text{phase-encoding}}$

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

$$k_y = \frac{\gamma}{2\pi} m g_y T_{\text{phase-encoding}}$$

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 ρ and T_2 -weighted protocols where TR is long to minimize T_1 effects

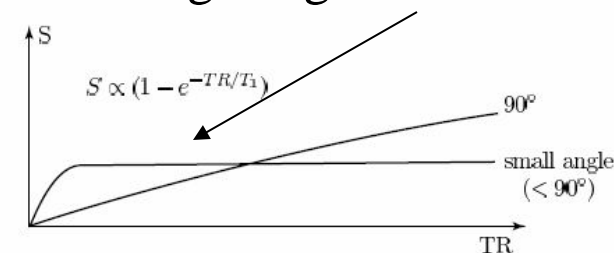
Gradient-Echo is an alternative technique designed to overcome these limitations:

- flip angle is less than 90° (usually 20° - 60°)
- no spin-echo due to absence of 180° rephasing pulse
- rephasing only by gradient reversal \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

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

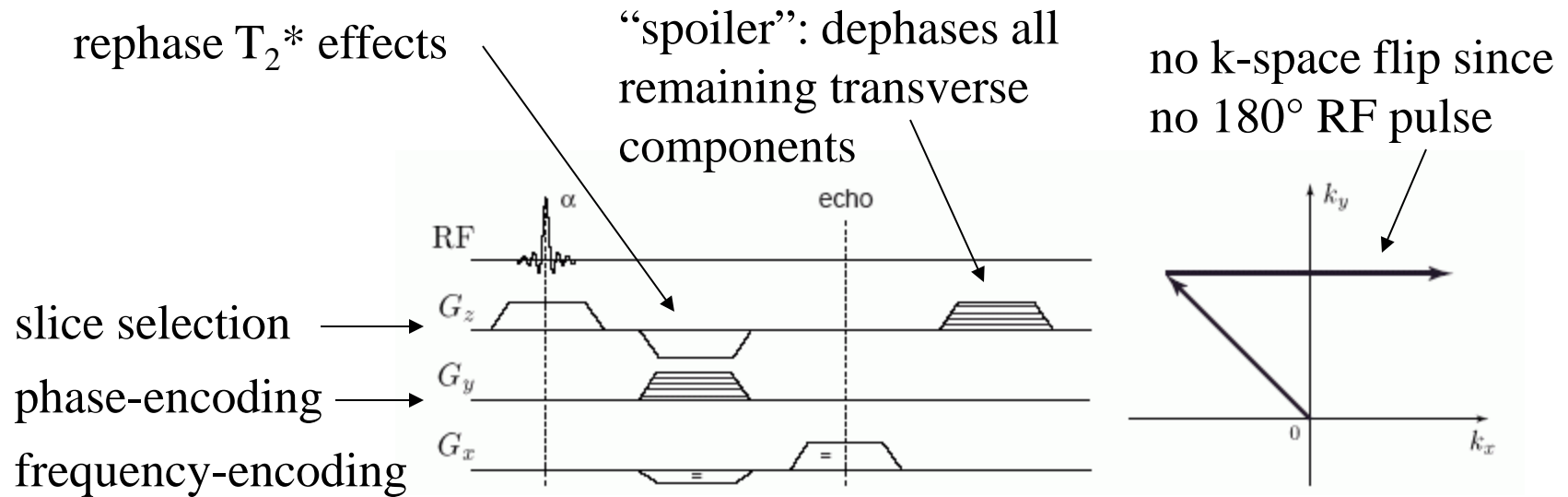


Spin-Echo compensates for dephasing
→ surrounding areas OK (with some geometric distortion)



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

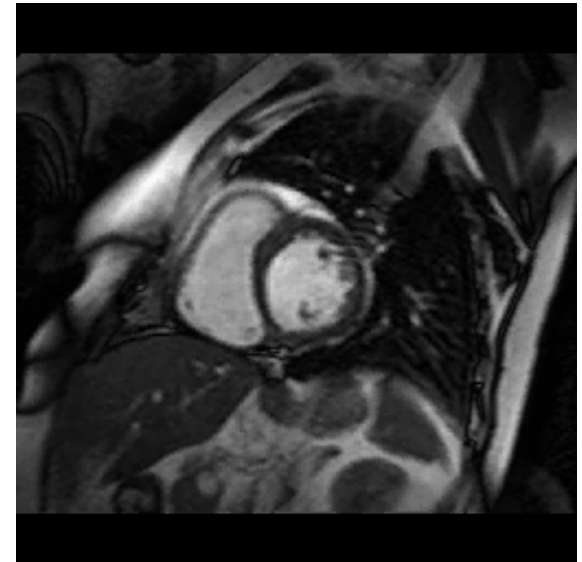
Gradient-Echo: FLASH



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, ...

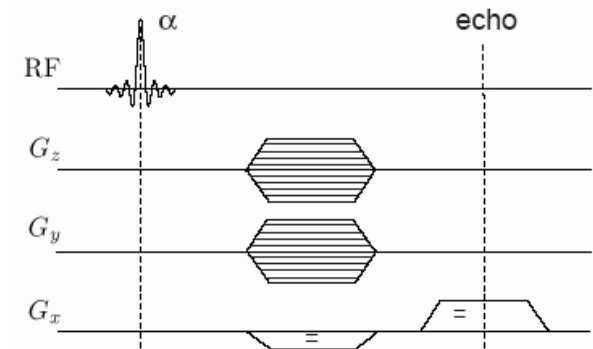
Requires a second phase-encoding gradient

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

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

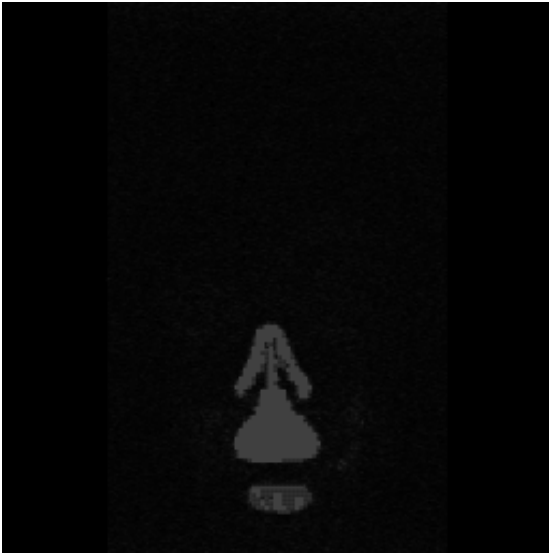
Acquisition time T_A determined by:

- 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)

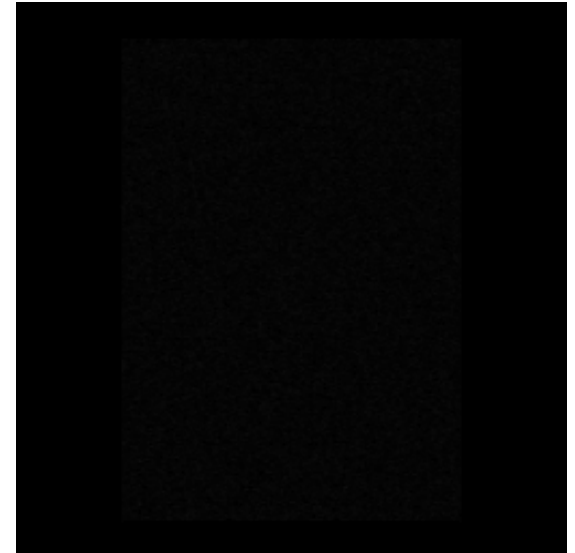
3D Clinical Imaging Example: Brain



Coronal cut



Sagittal cut



Axial cut

Assume a $32 \times 256 \times 256$ volume:

- $N_{ac}=1, N_{sl}=32, N_{ph}=256$
- with Spin-Echo, $TR=2000$ ms $\rightarrow TA_{3D} = 4$ hours (infeasible)
- with Gradient-Echo, $TR=40$ ms $\rightarrow TA_{3D} = 6$ minutes (acceptable)

Fast Imaging Sequences

Recall acquisition time equation:

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

Could get faster by

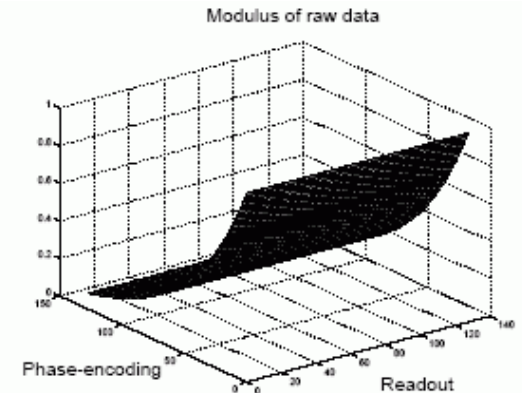
- decreasing TR (GE (gradient echo) vs. SE (spin echo) sequences)
- decreasing N_{ac} (truncated or half Fourier imaging)
- or...generate and acquire multiple echos in one excitation (ETL = Echo Train Length)

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



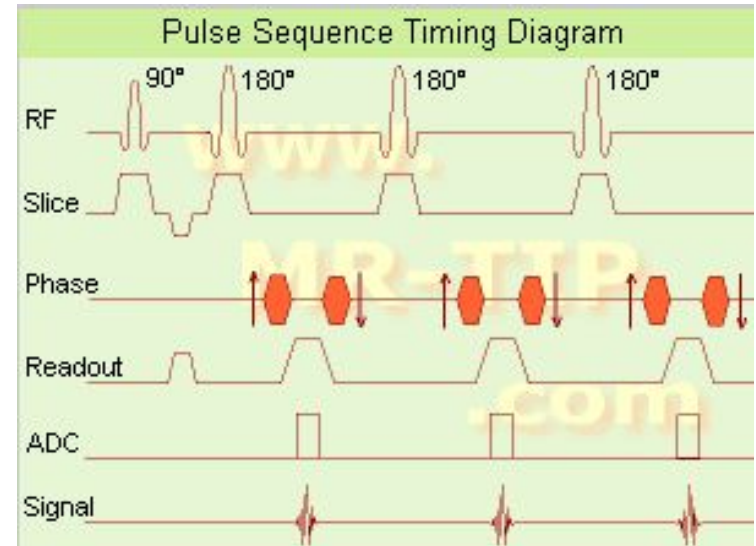
Fast Sequences: TurboSE

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:

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

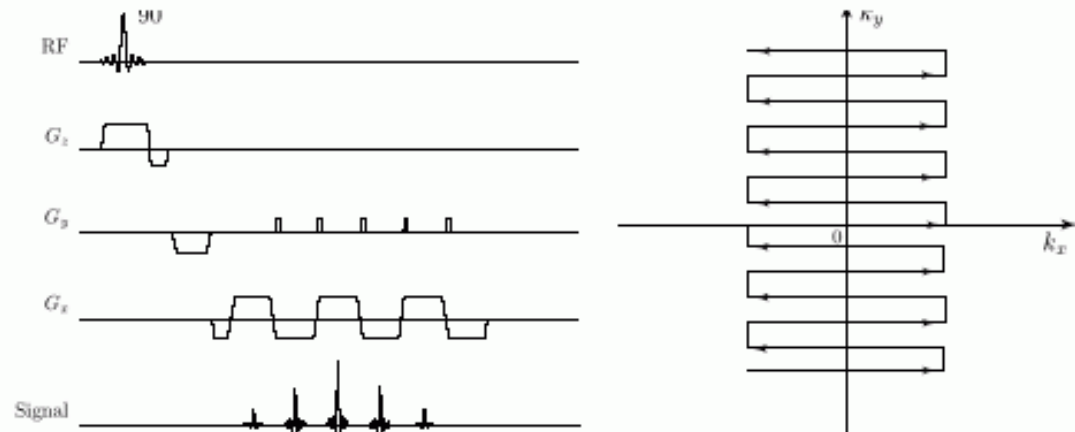
Fast Sequences: HASTE and EPI

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

- all echos are generated within one excitation $\rightarrow \text{ETL} = N_{\text{ph}}$
- and thus: $\text{TA} = \text{TR}$ (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
= 128×128
- 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 between prostate tissue (white outline) and adjacent peripheral structures

MRI Tissue Characterization

(liver)

T1-weighted GE

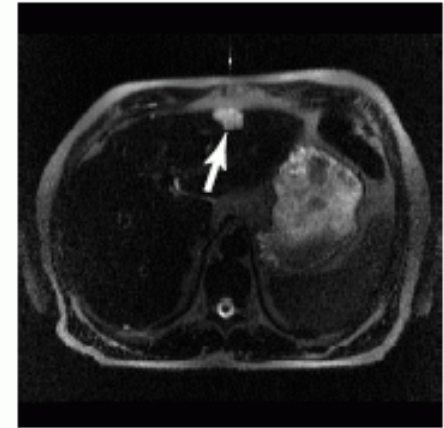
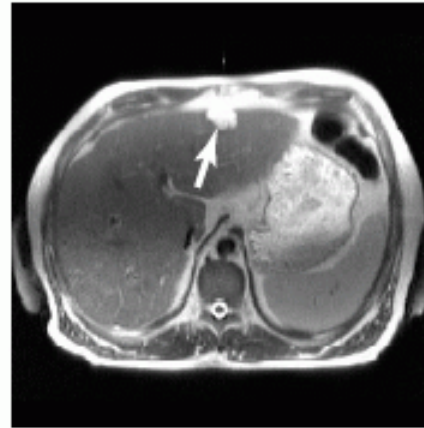
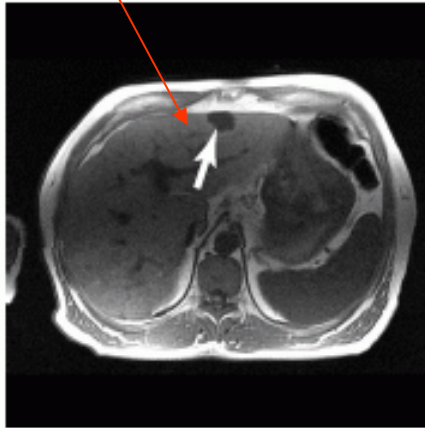
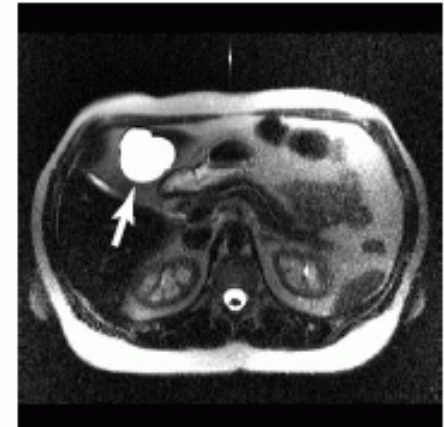
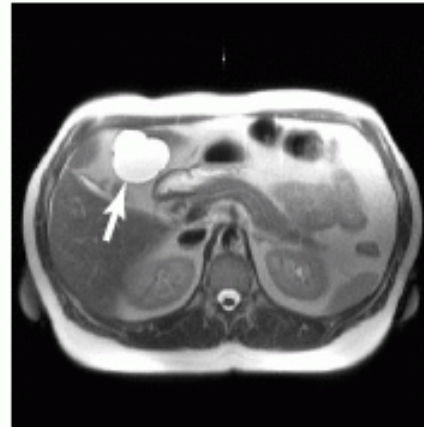
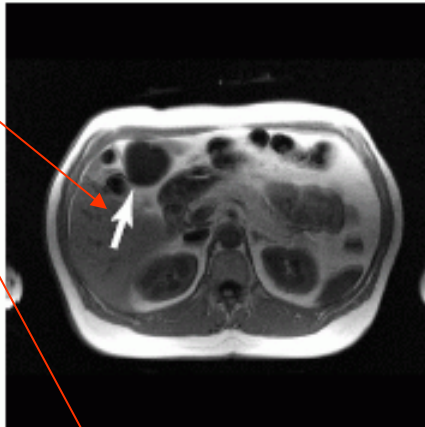
T2-weighted (HASTE)

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 delivers part of the oxygen 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

Known 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 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

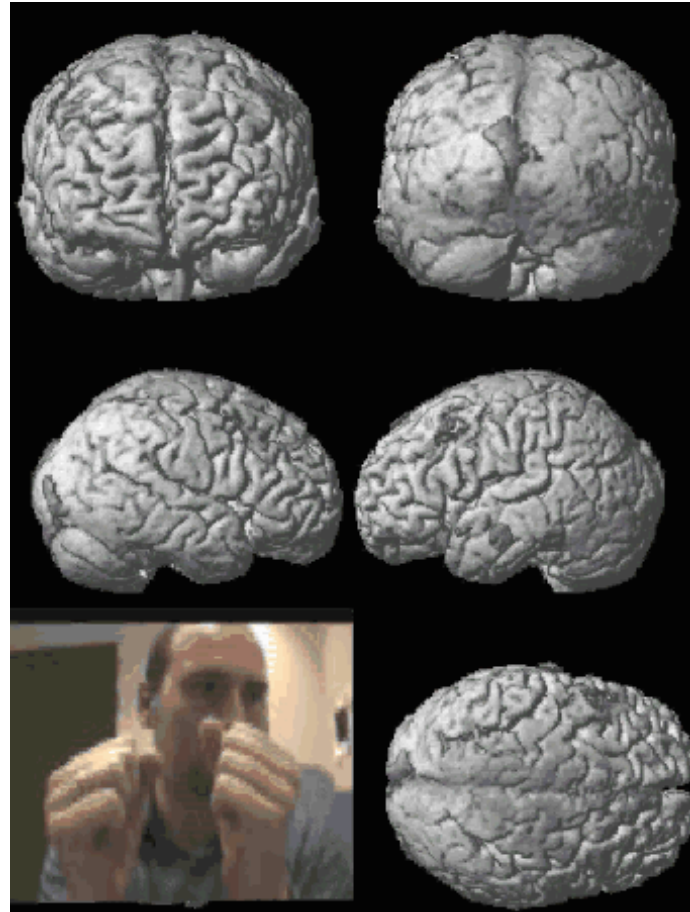
Many repetitions needed

- 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

fMRI Applications

Get insight into:

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



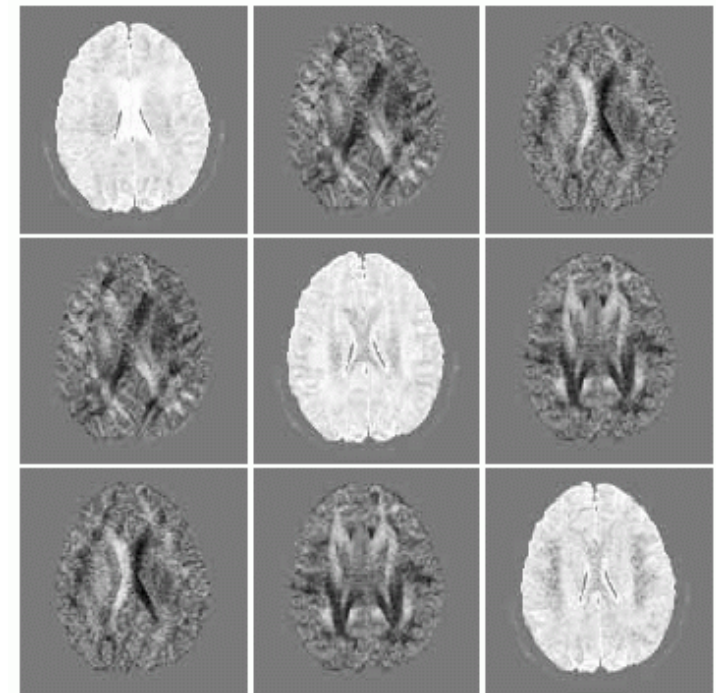
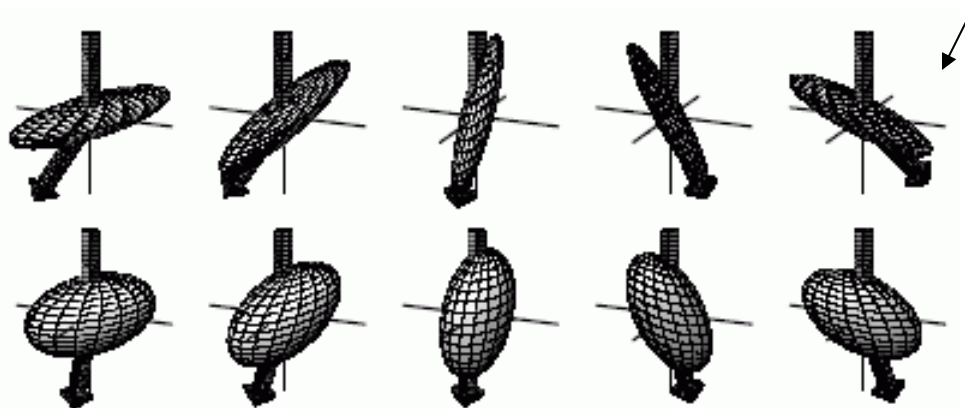
Diffusion Tensor Imaging (DTI)

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.

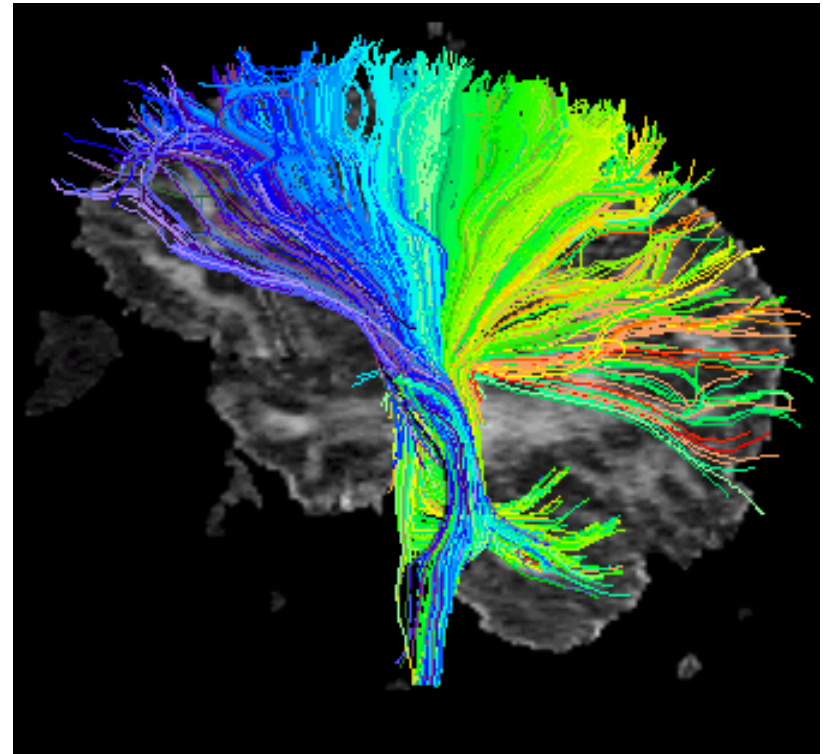
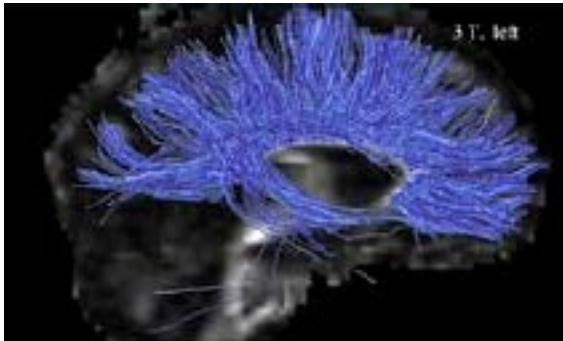
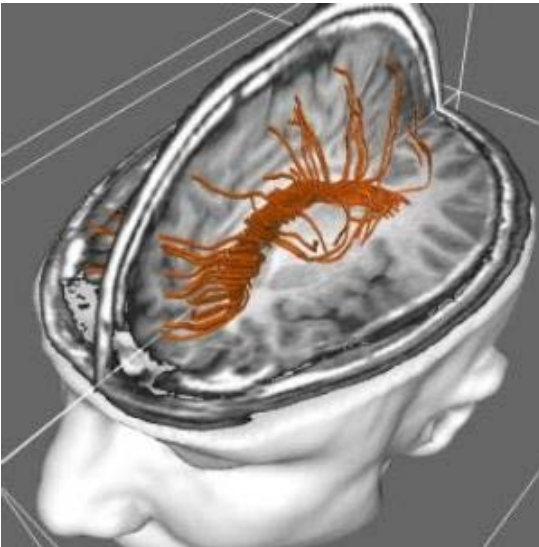
Acquisition and analysis

- 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



Diffusion Tensor Imaging (DTI)

