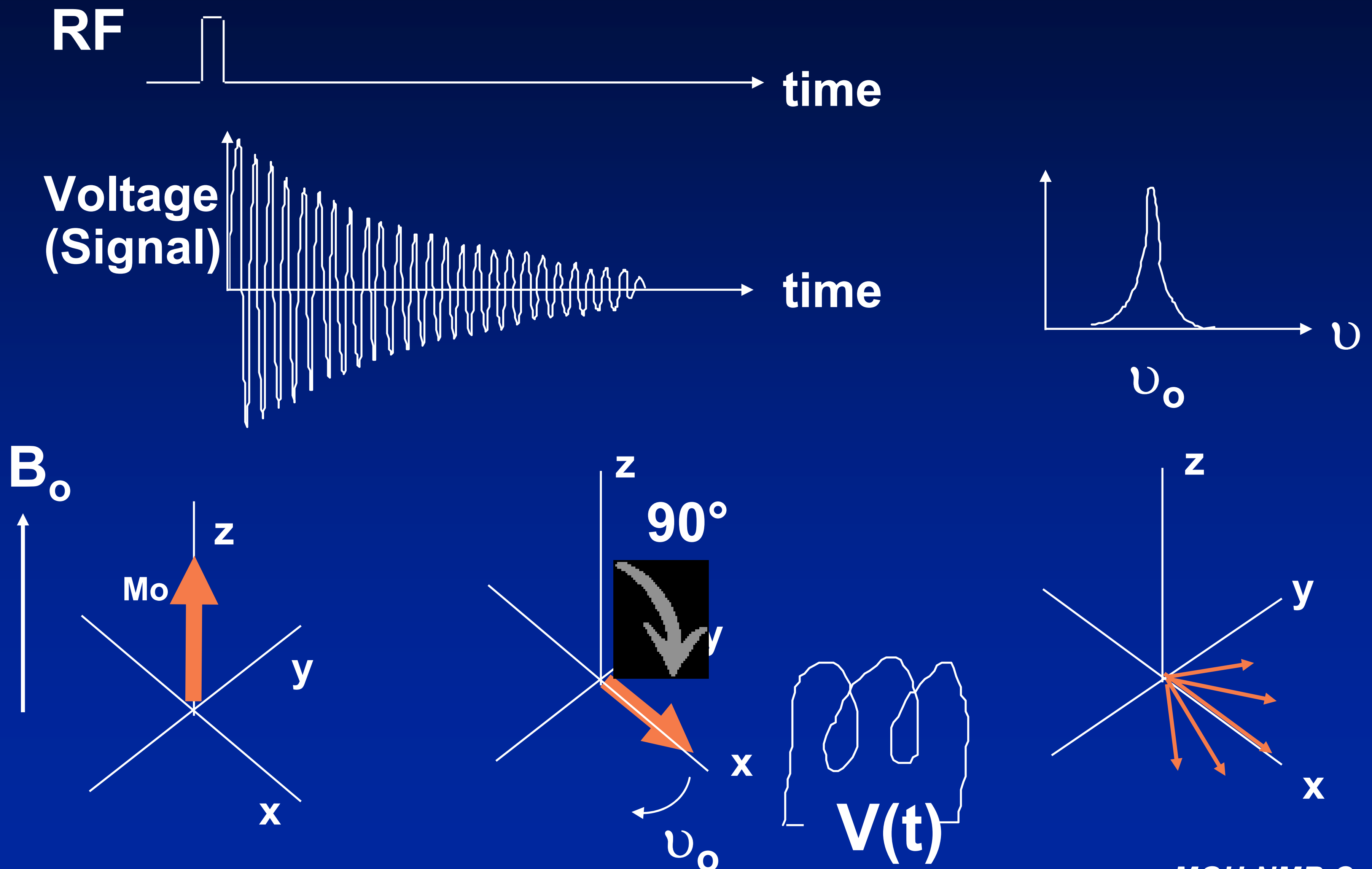
Basic MR image encoding

MGH-NMR Center

HST.583: Functional Magnetic Resonance Imaging: Data Acquisition and Analysis
Harvard-MIT Division of Health Sciences and Technology
Dr. Larry Wald

Review: the NMR Signal



Three Steps in MR:

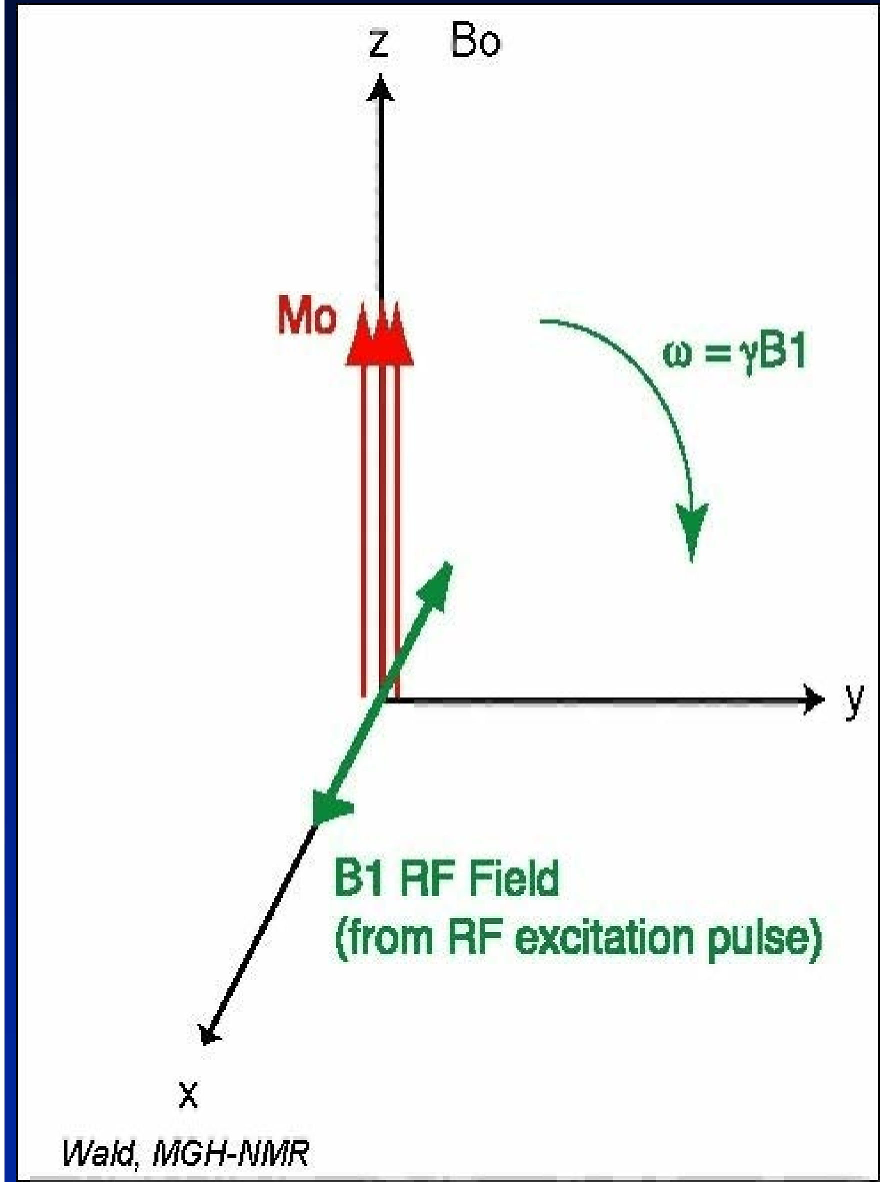
0) Equilibrium (magnetization points along Bo)

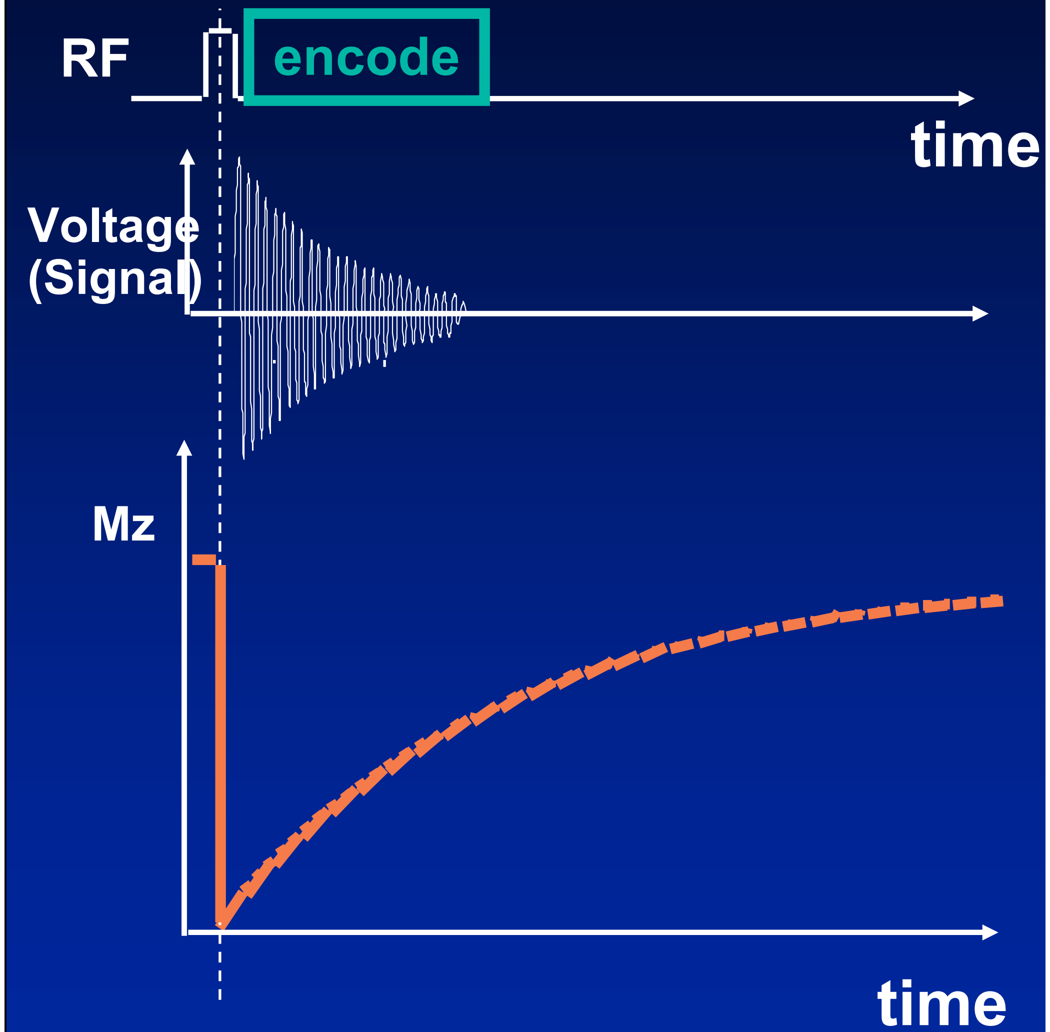
1) RF Excitation (tip magn. away from equil.)

2) Precession induces signal,
dephasing (timescale = T2, T2*).

3) Return to equilibrium (timescale = T1).

Magnetization vector durning MR





Three places in process to make a measurement (image)

0) Equilibrium (magnetization points along Bo)

1) RF Excitation (tip magn. away from equil.)

← proton density weighting

2) Precession induces signal, allow to dephase for time TE.

← T2 or T2* weighting

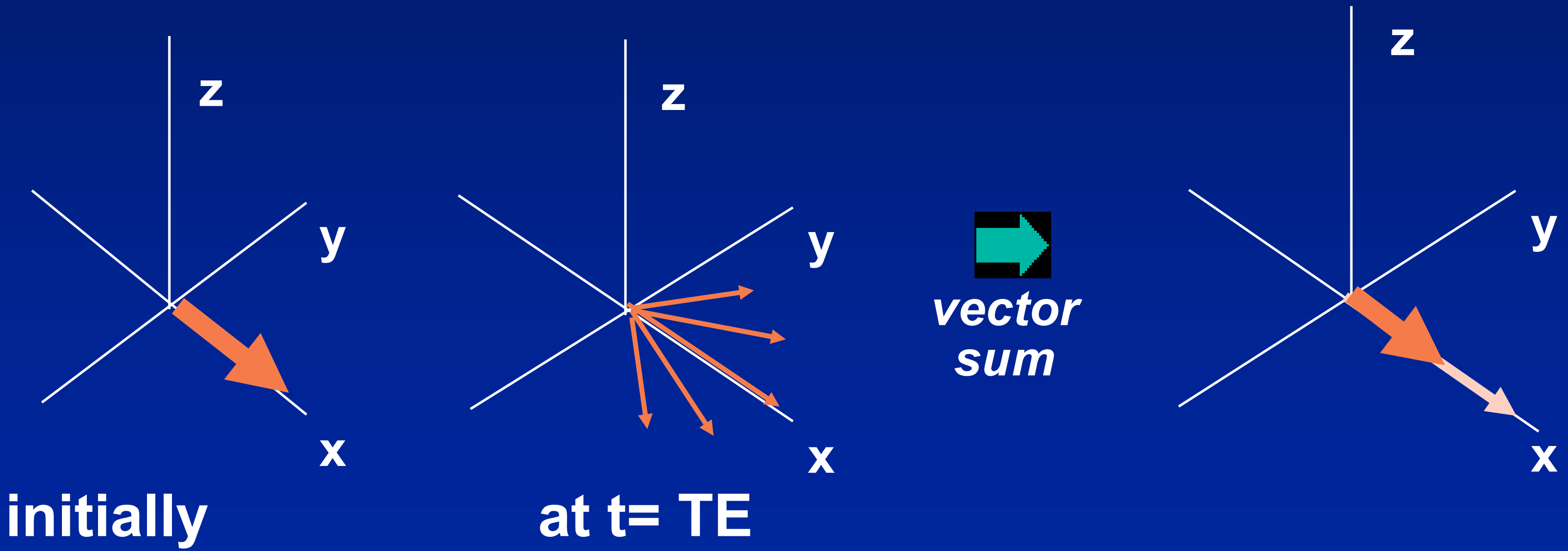
3) Return to equilibrium (timescale =T1).

← T1 Weighting

T2*-Weighting

Wait time <u>TE</u> after excitation before measuring M.

Shorter T2* spins have dephased



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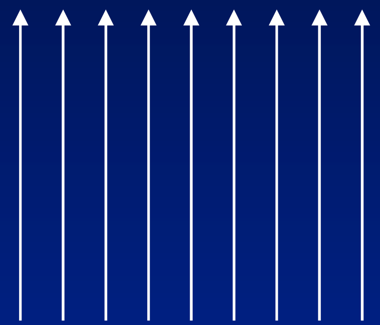
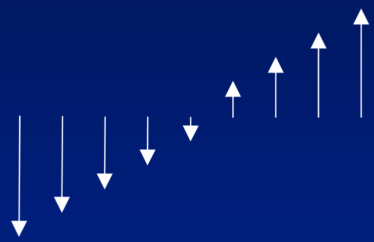
Aside: Magnetic field gradient

B_o



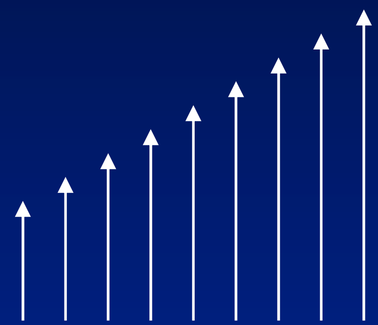
Uniform magnet

$G_x\, x$

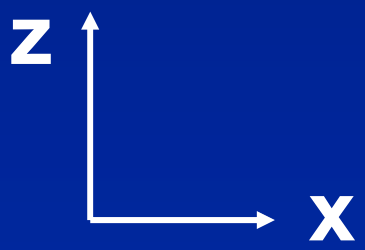


Field from gradient coils

$B_o + G_x\, x$



Total field

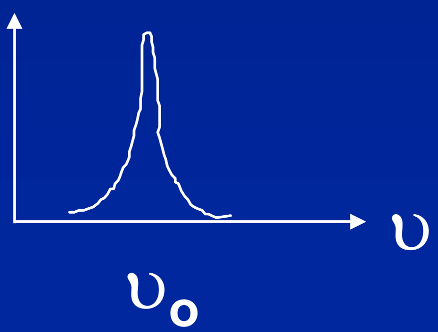


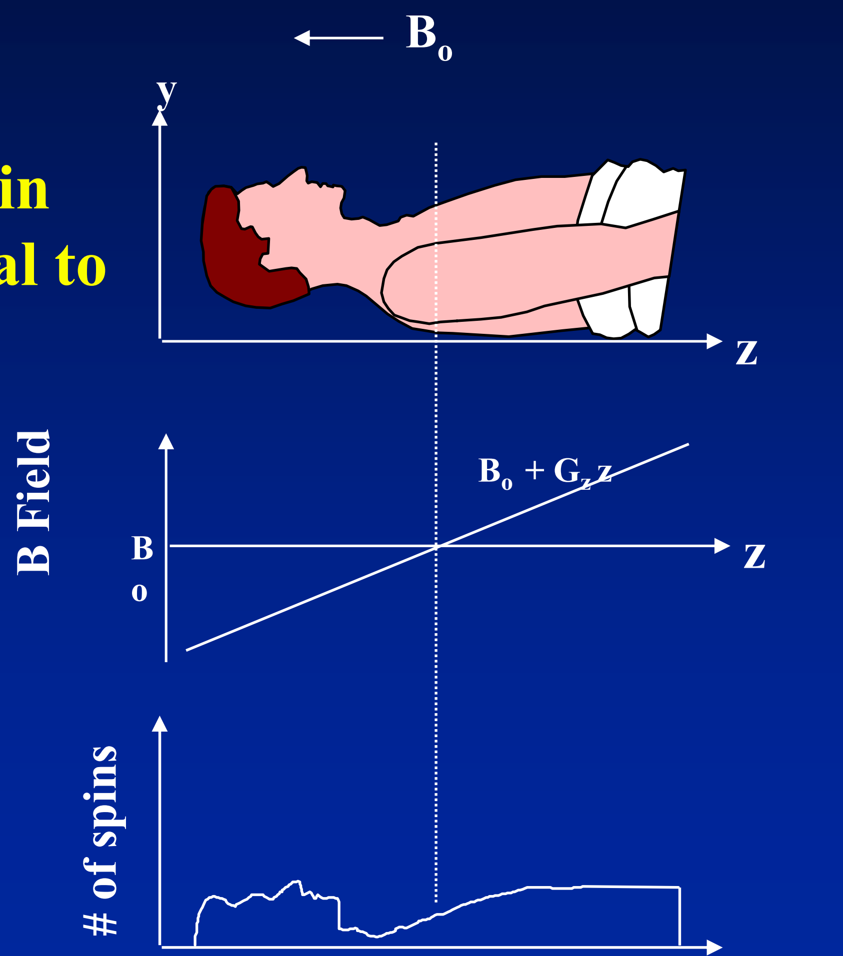
$$G_x \quad \partial B_z / \partial x$$

A gradient causes a spread of frequencies

MR frequency of the protons in a given location is proportional to the local applied field.

$$v = \gamma B_{TOT} = \gamma(B_o + G_z z)$$





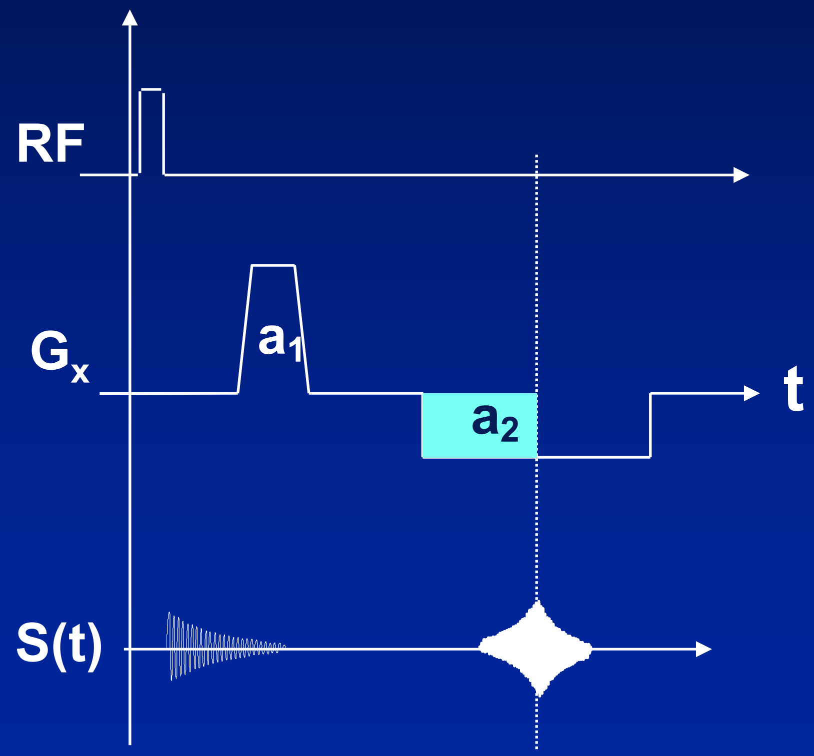
A gradient causes dephasing

I caused it, I can reverse it…
Gradient echo

$$\upsilon = \gamma B_{TOT} = \gamma\ B_o + G_z\ z$$

$$\Delta\upsilon = \gamma\Delta B_{TOT} = \gamma\ G_z\ z$$

$$\Delta\theta = \Delta\upsilon\ \tau = \gamma\ G_z\ z\ \tau$$

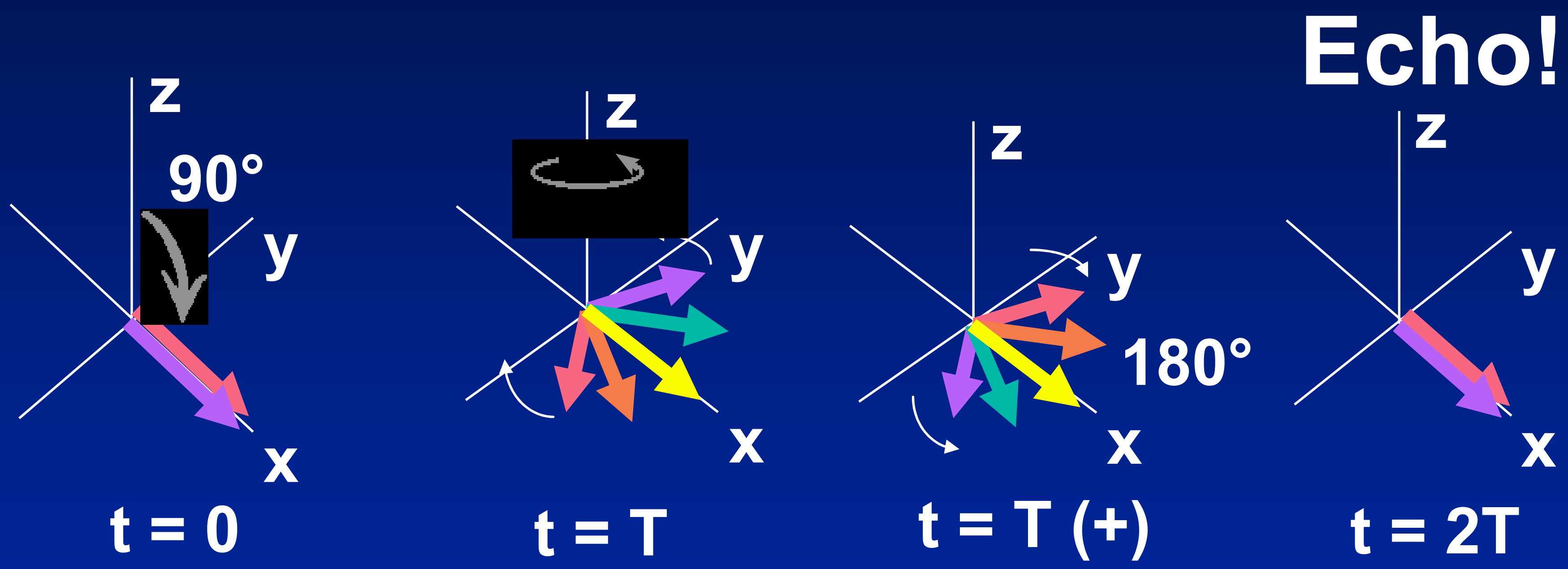


Gratuitous manipulation…
(?)

What happens if the spin moves?

Spin Echo

Some dephasing can be refocused because its due to static fields.



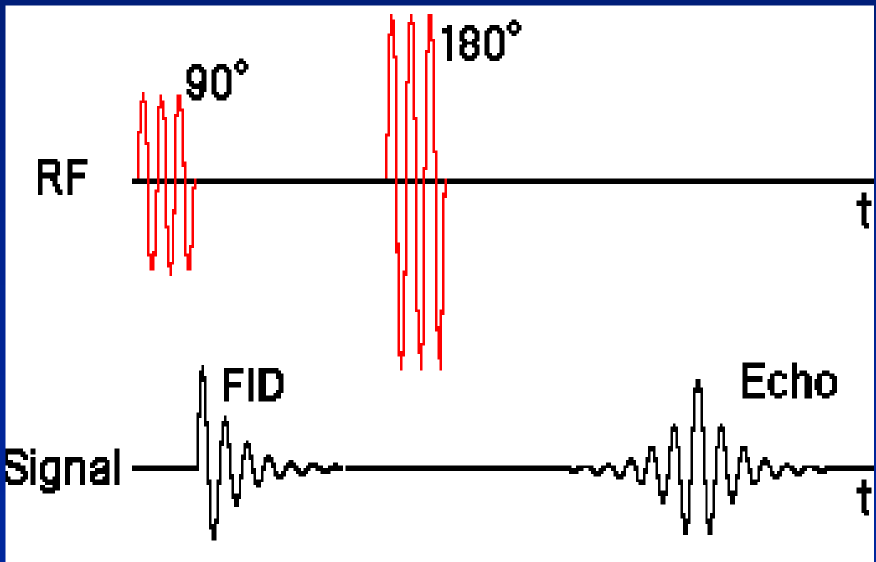
Blue/Green arrows precesses faster due to local field inhomogeneity than red/orange arrow

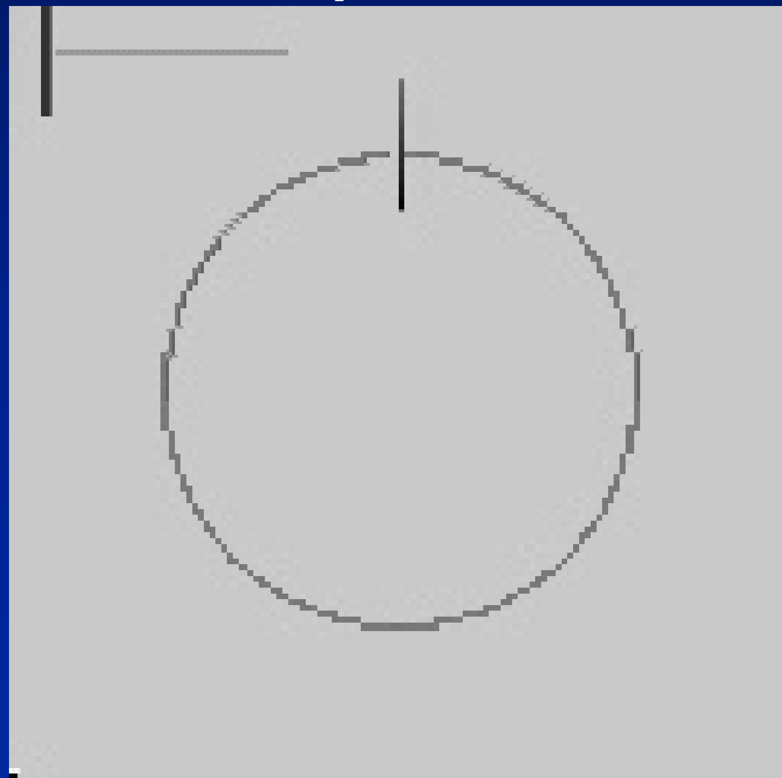
Spin Echo

180° pulse only helps cancel static inhomogeneity

The "runners" can have <u>static</u> speed distribution.

If a runner trips, he will not make it back in phase with the others.





Other contrast for MRI

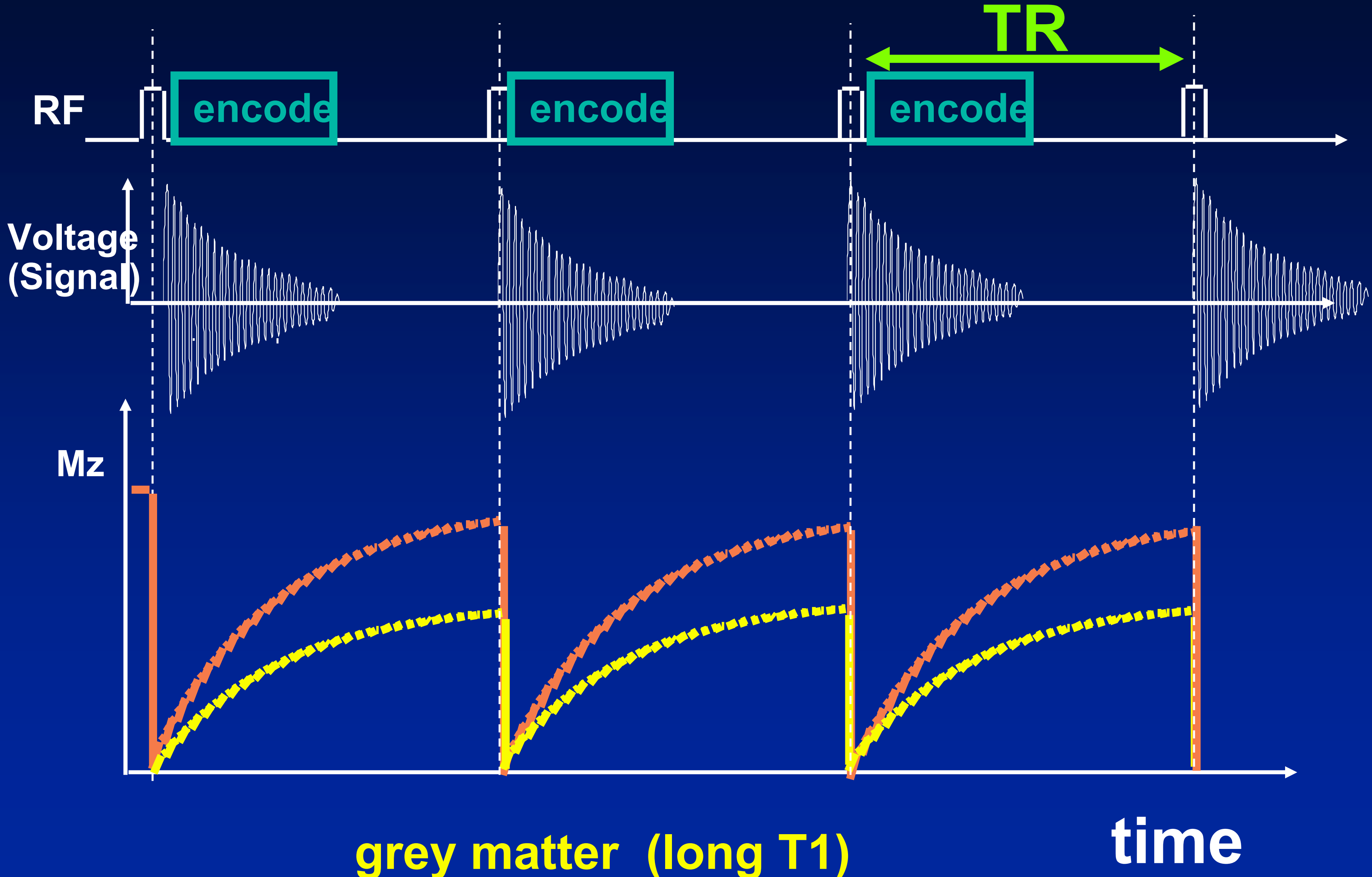
In brain: (gray/white/CSF/fat)
Proton density differ ~ 20%
T1 relaxation differ ~ 2000%

How to exploit for imaging?

Answer:

Vary repetition rate - TR

T1 weighting in MRI



grey matter (long T1)
white matter (short T1)

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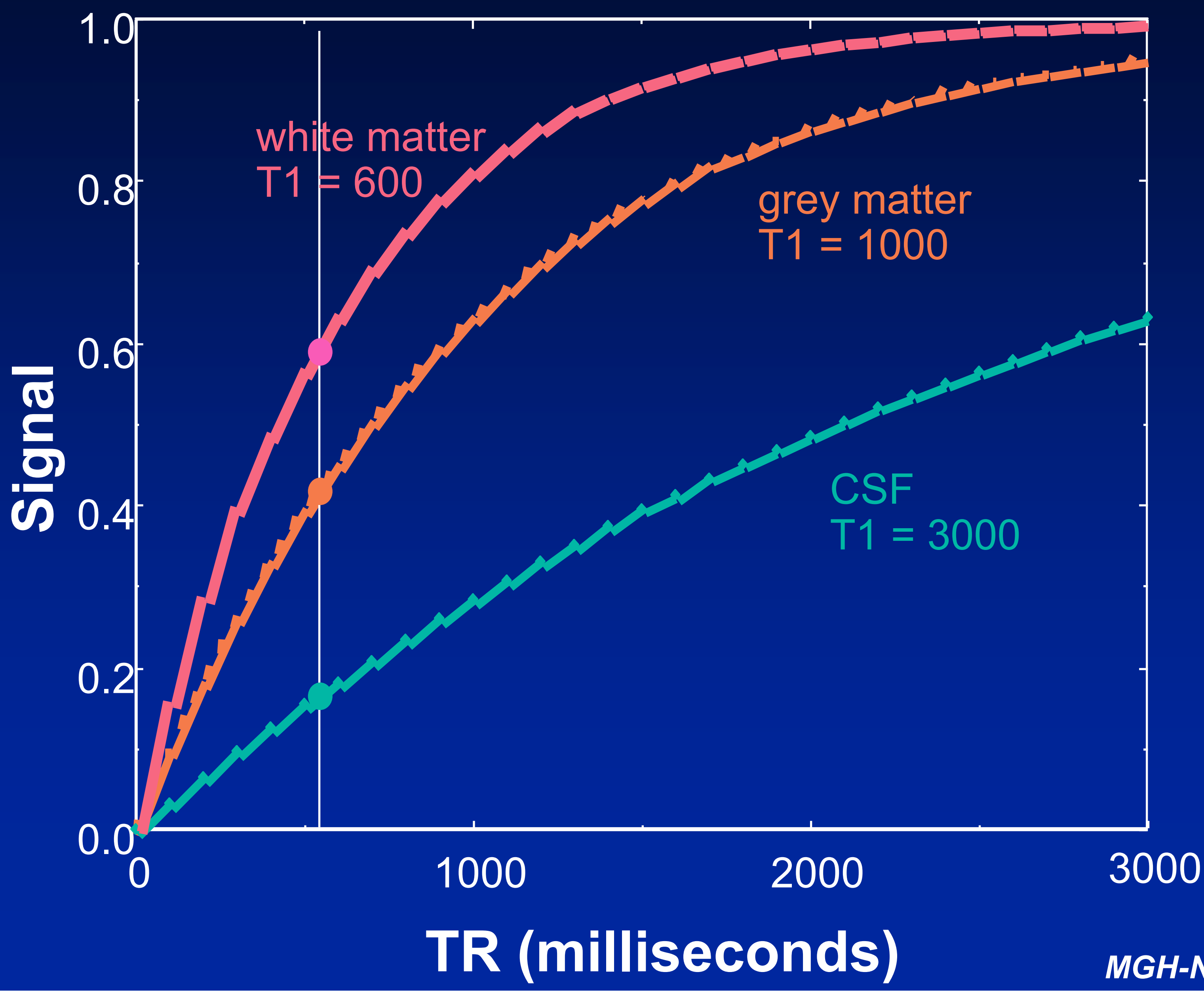
T1-Weighting



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T1-Weighting

Very long TR, Signal ~ Proton Density

Shortening TR $\rightarrow$ long T1 darker

"Best" T1 contrast, TR near T1

Contrast comes at expense of Signal: throw away some magnetization

Summarizing Contrast

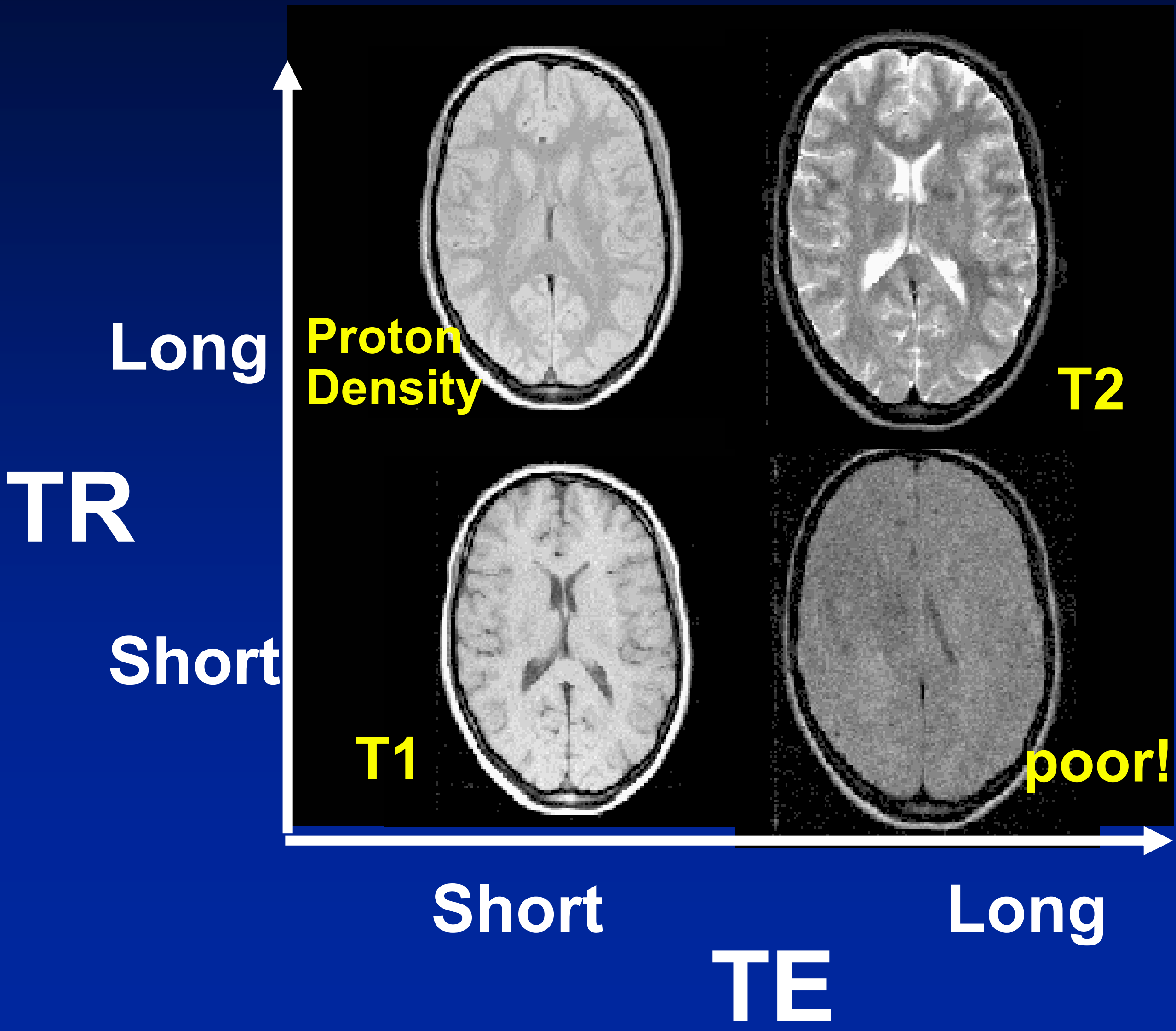
Two knobs: TE and TR

Effects tend to compete, most tissue with long T1 also has long T2 and contrast goes opposite ways.

T1-weighting: short T1 is bright

T2-weighting: long T2 bright

Image contrast through choices of TR, TE



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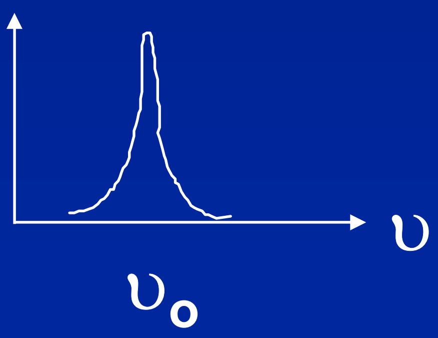
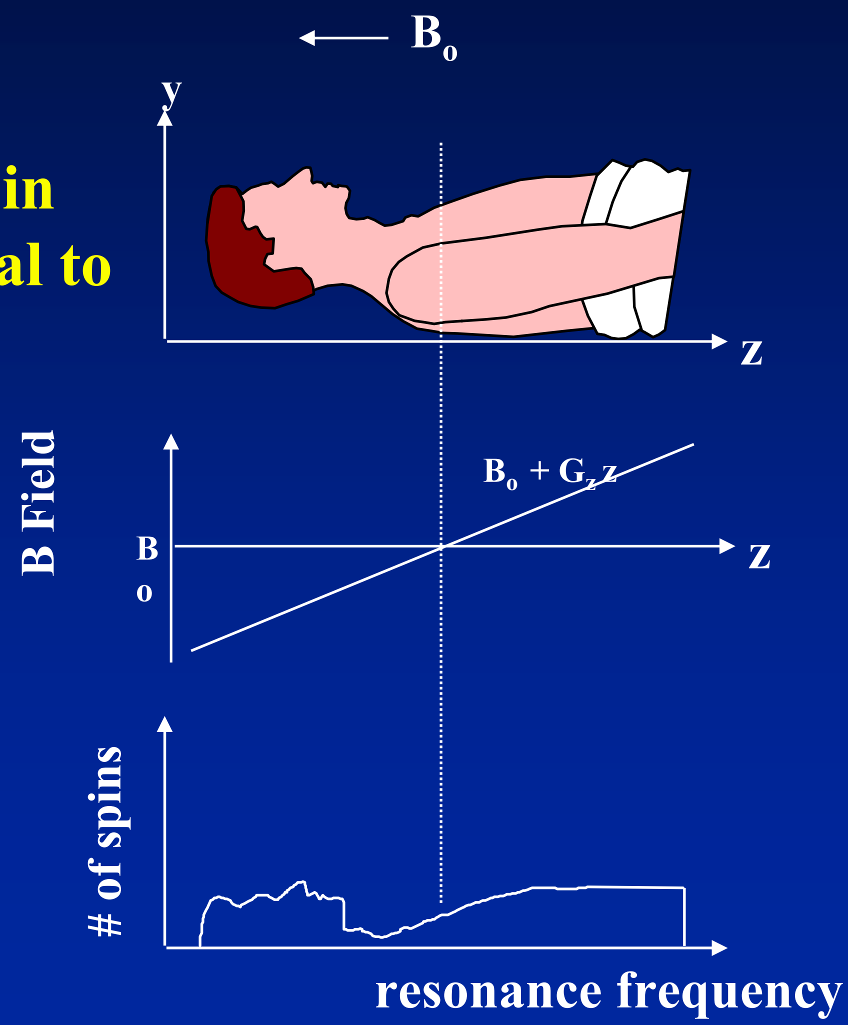
1D projection image

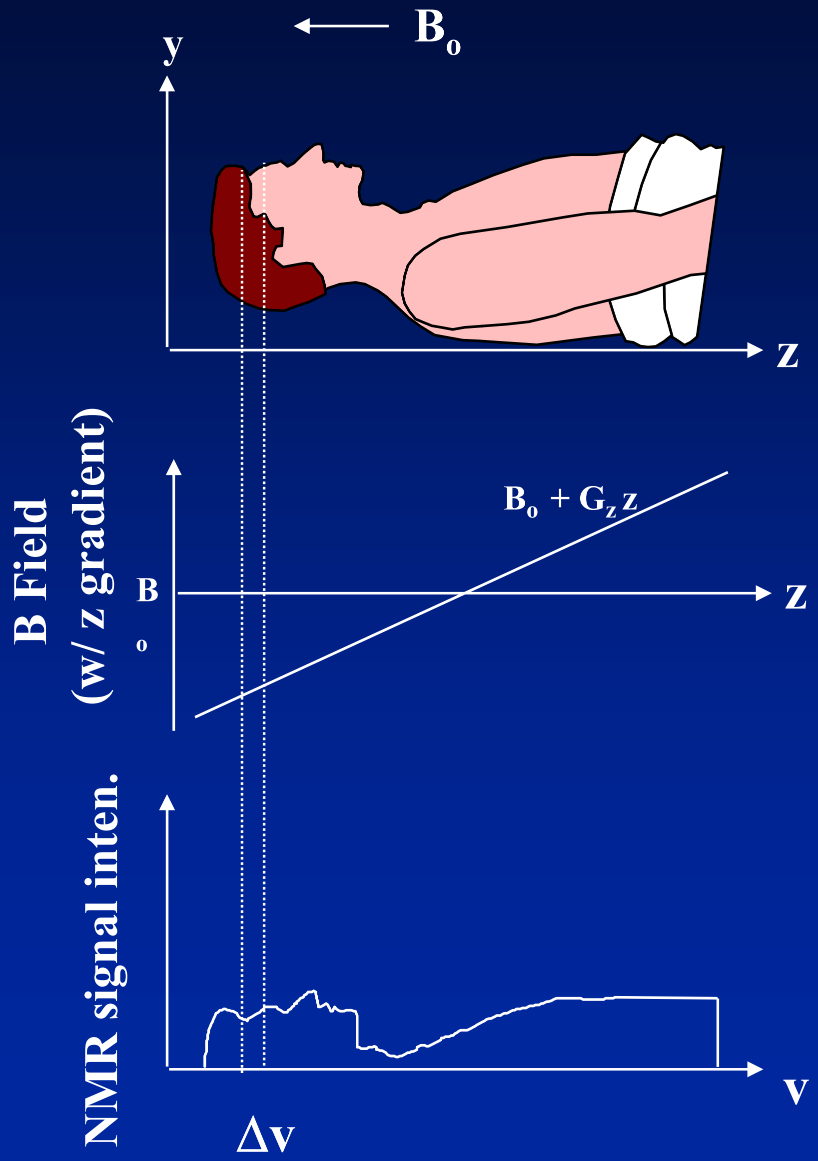
MR frequency of the protons in a given location is proportional to the local applied field.

$$v = \gamma B_{TOT} = \gamma(B_o + G_z z)$$

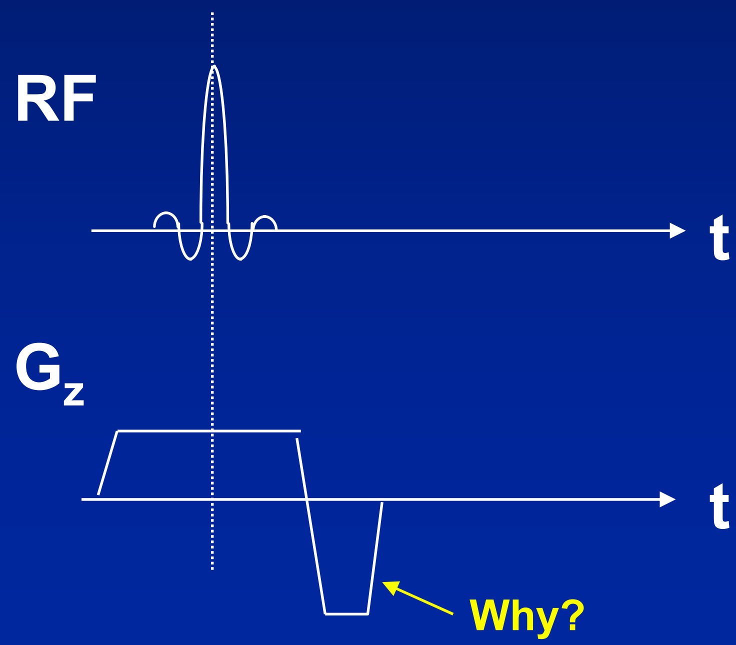




Step one: excite a slice

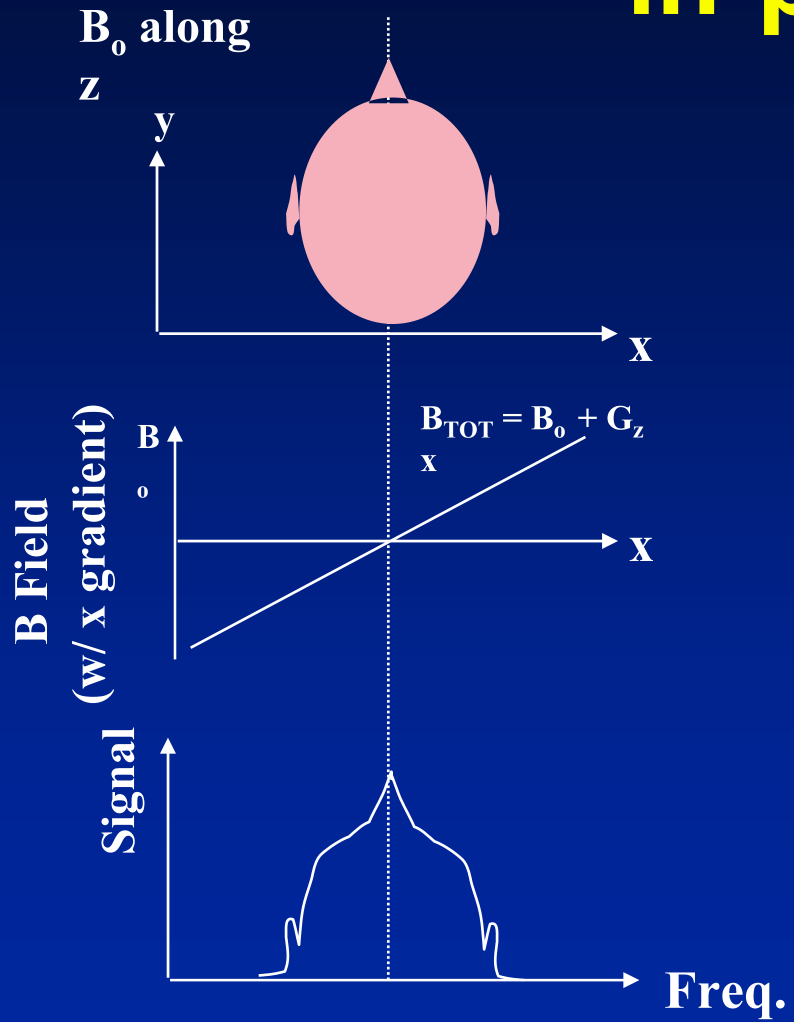


While the grad. is on, excite only band of frequencies.

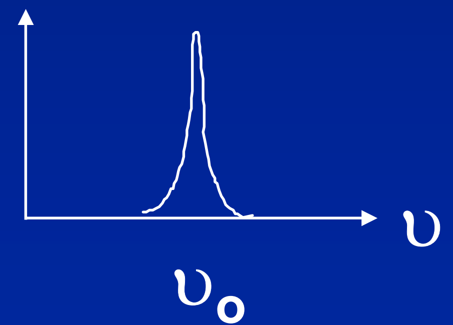


Step two: encode spatial info. in-plane

“Frequency encoding”

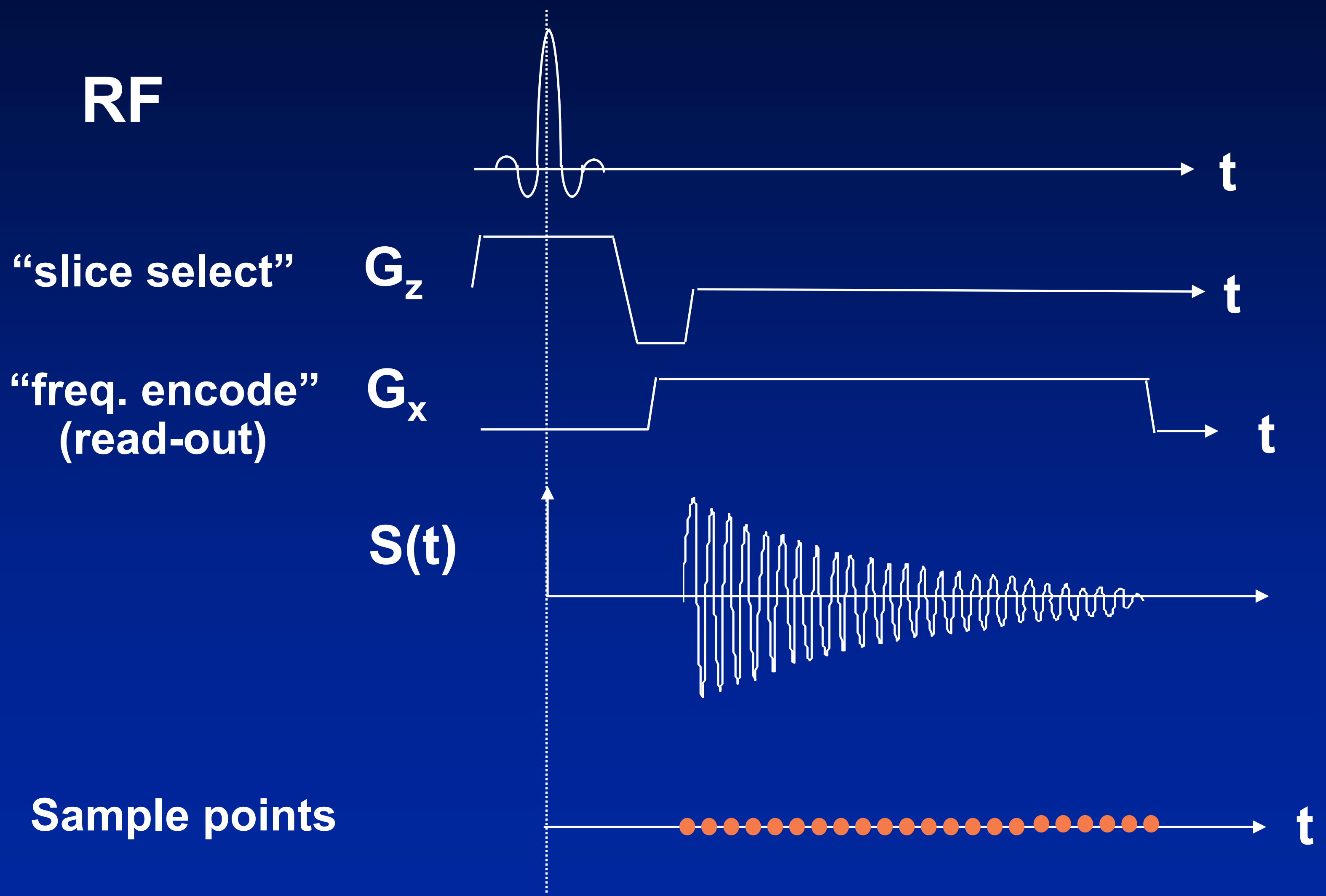


with gradient

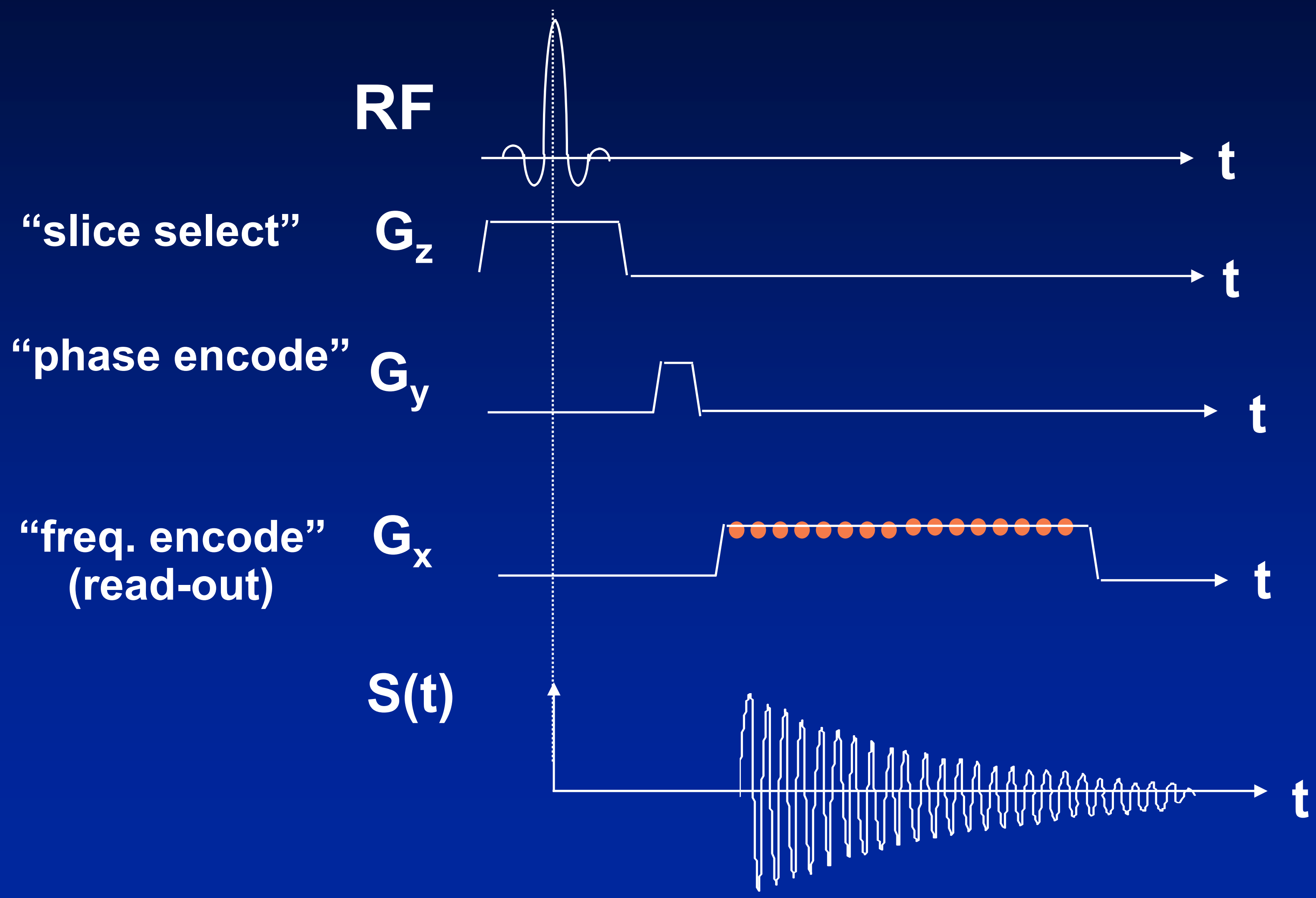


without gradient

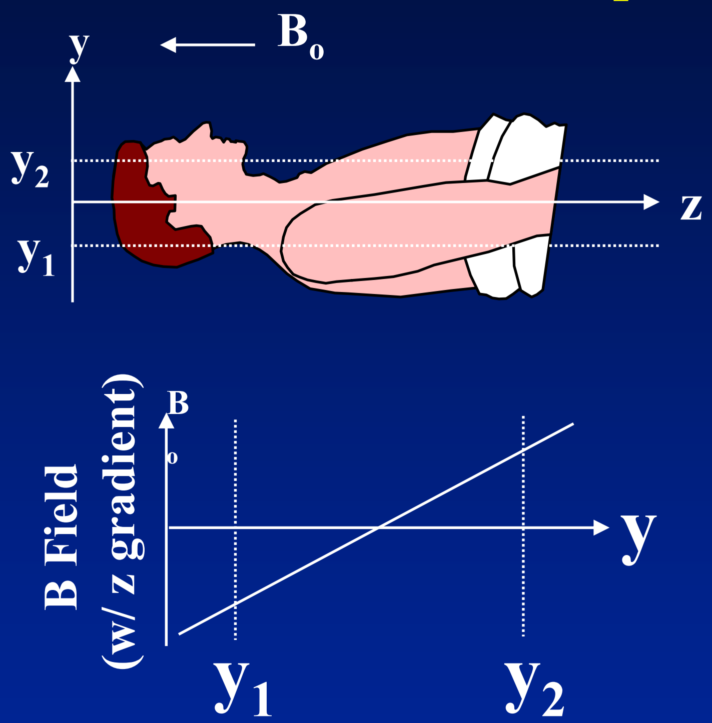
‘Pulse sequence’ so far

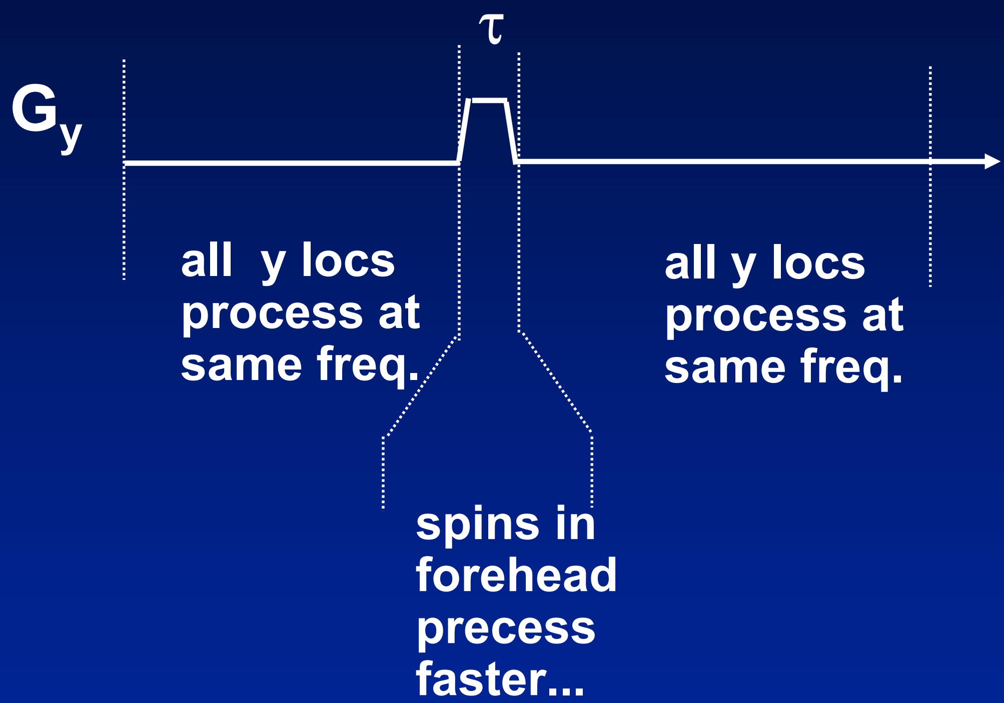


“Phase encoding”



How does blipping on a grad. encode spatial info?

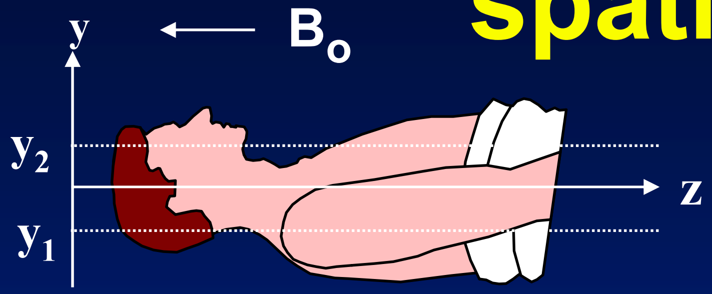




$$\upsilon(y) = \gamma B_{TOT} = \gamma B_o \Delta y G_y$$

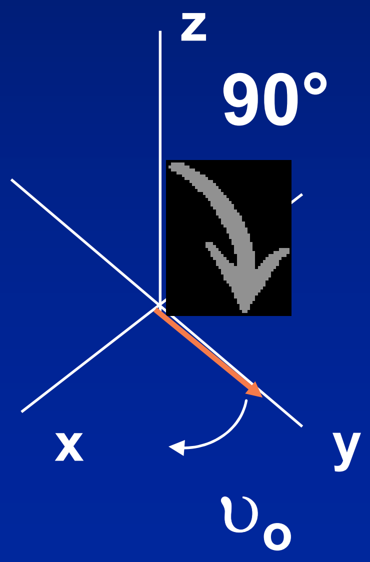
$$\theta(y) = \upsilon(y)\tau = \gamma B_o \Delta y (G_y \tau)$$

How does blipping on a grad. encode spatial info?

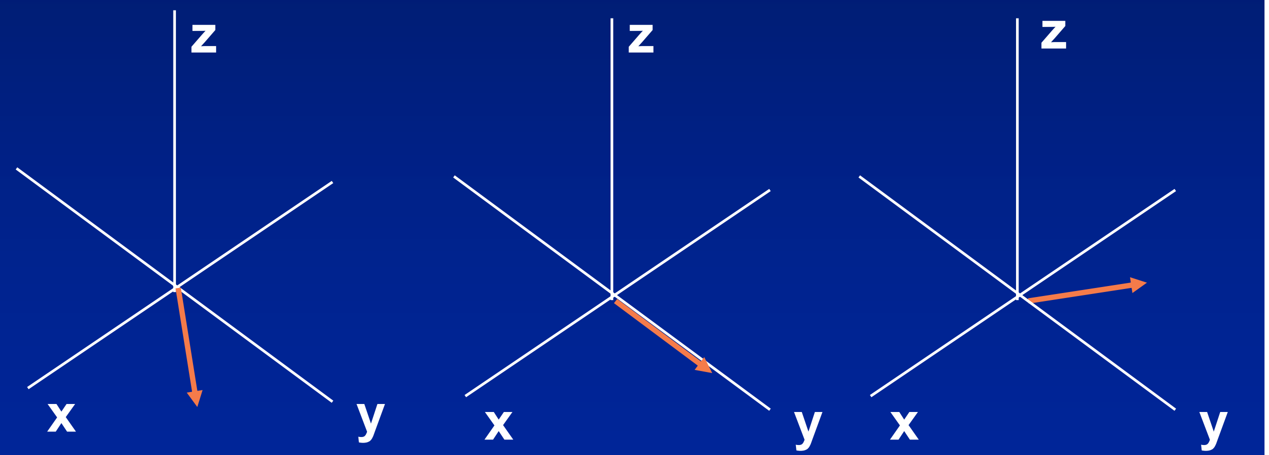


$$\theta\,(y) = \upsilon(y)\,\tau \;\; = \gamma\, B_o\, \Delta y\, (G_y\, \tau)$$

after RF

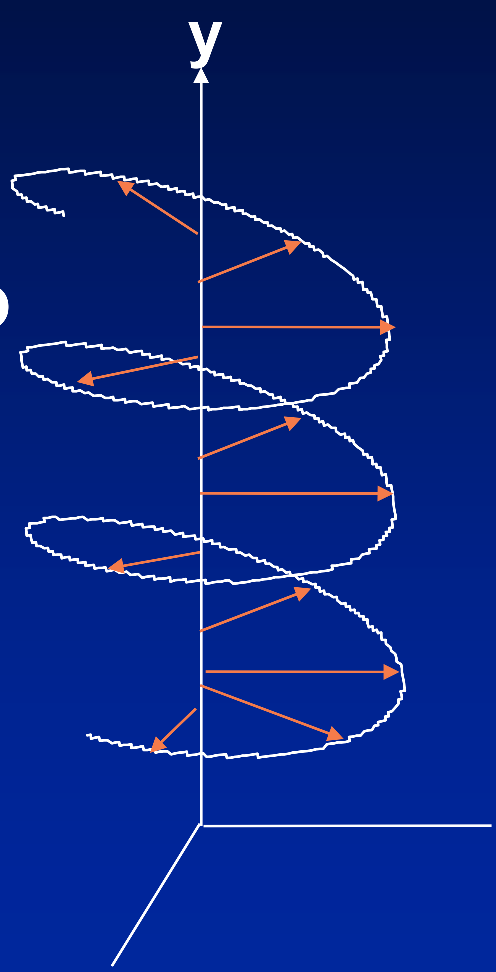


After the blipped y gradient...



position y_1 **position 0** **position y_2**

How does blipping on a grad. encode spatial info?

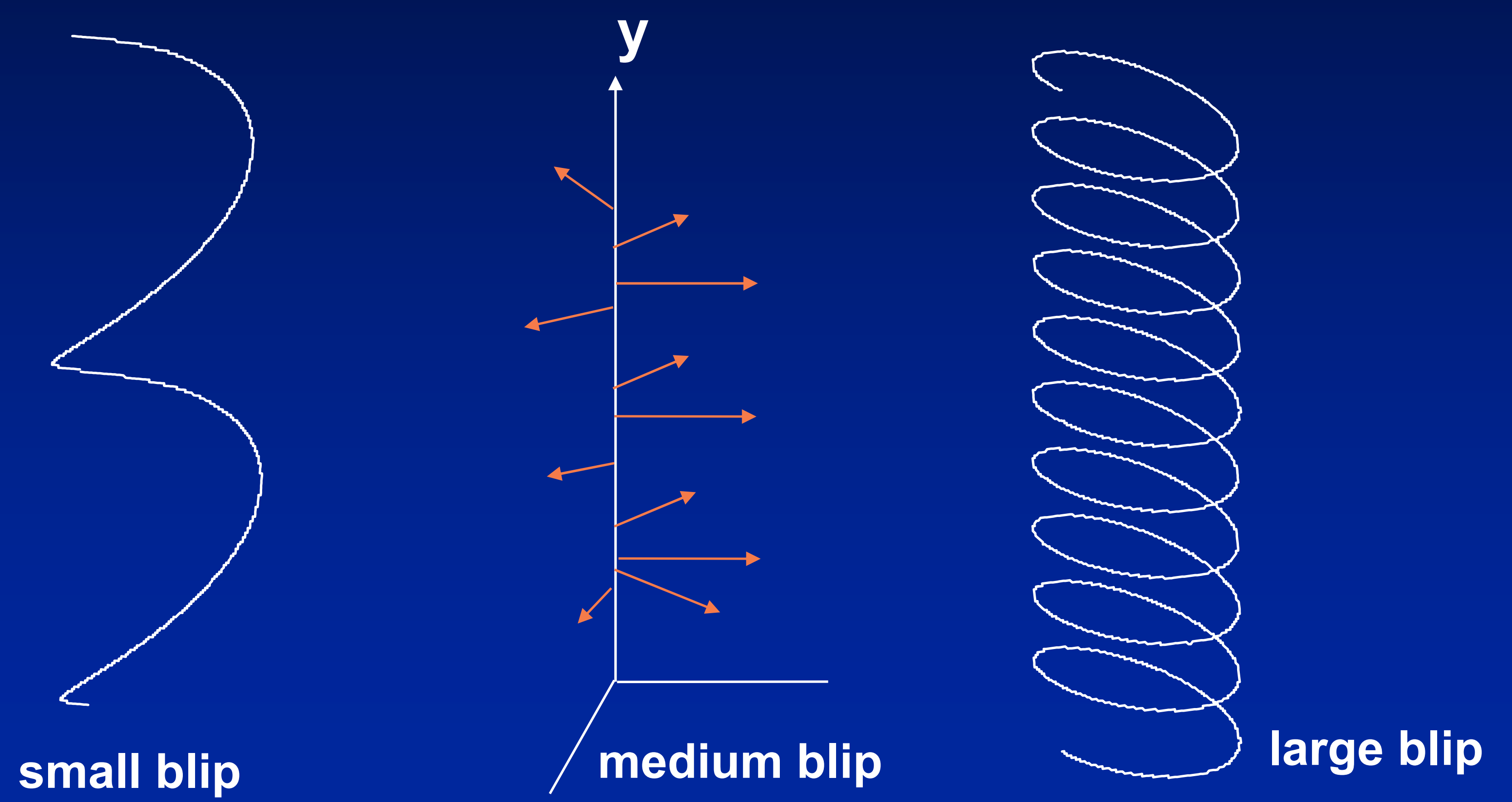


The magnetization vector in the xy plane is wound into a helix directed along y axis.

Phases are ‘locked in’ once the blip is over.

The bigger the gradient blip area, the tighter the helix

$$\theta(y) = \upsilon(y)\,\tau = \gamma\, B_o\, \Delta y\, (G_y\, \tau)$$



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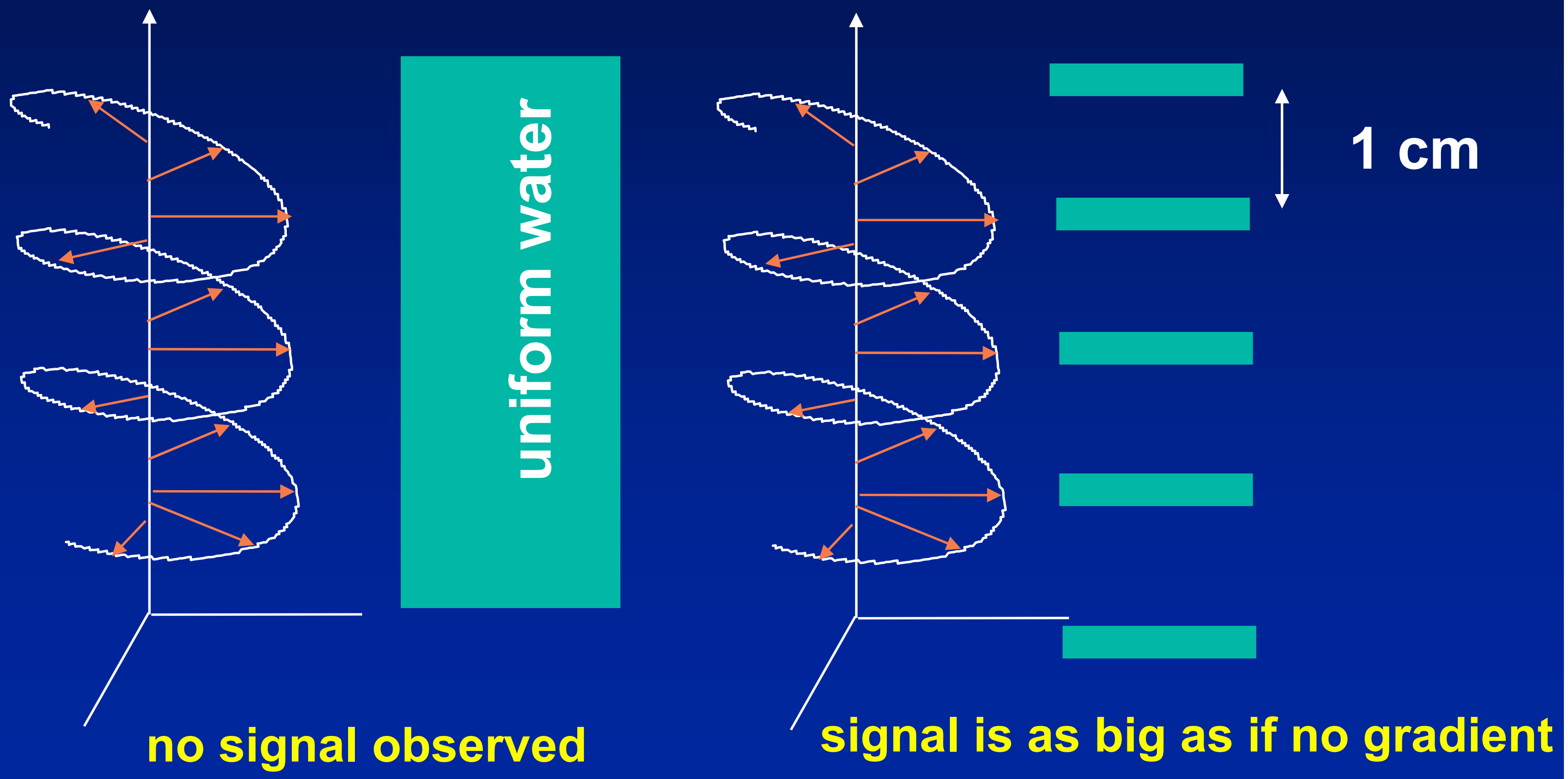
What have you measured?

Consider 2 samples:



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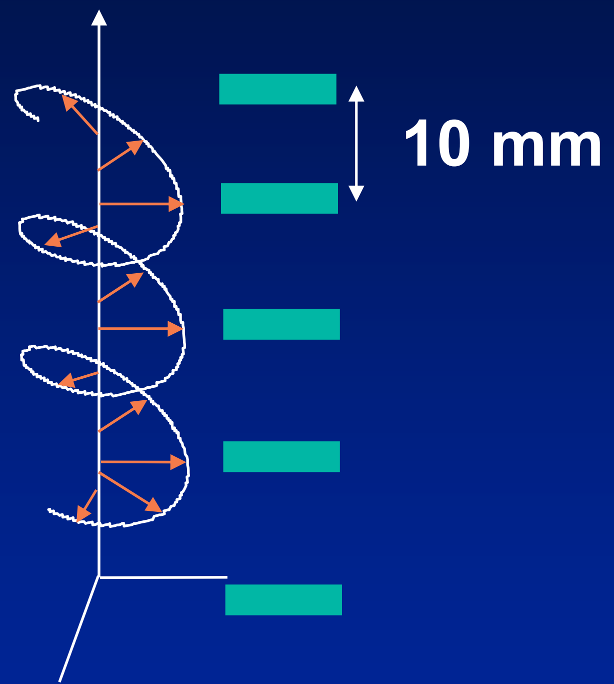
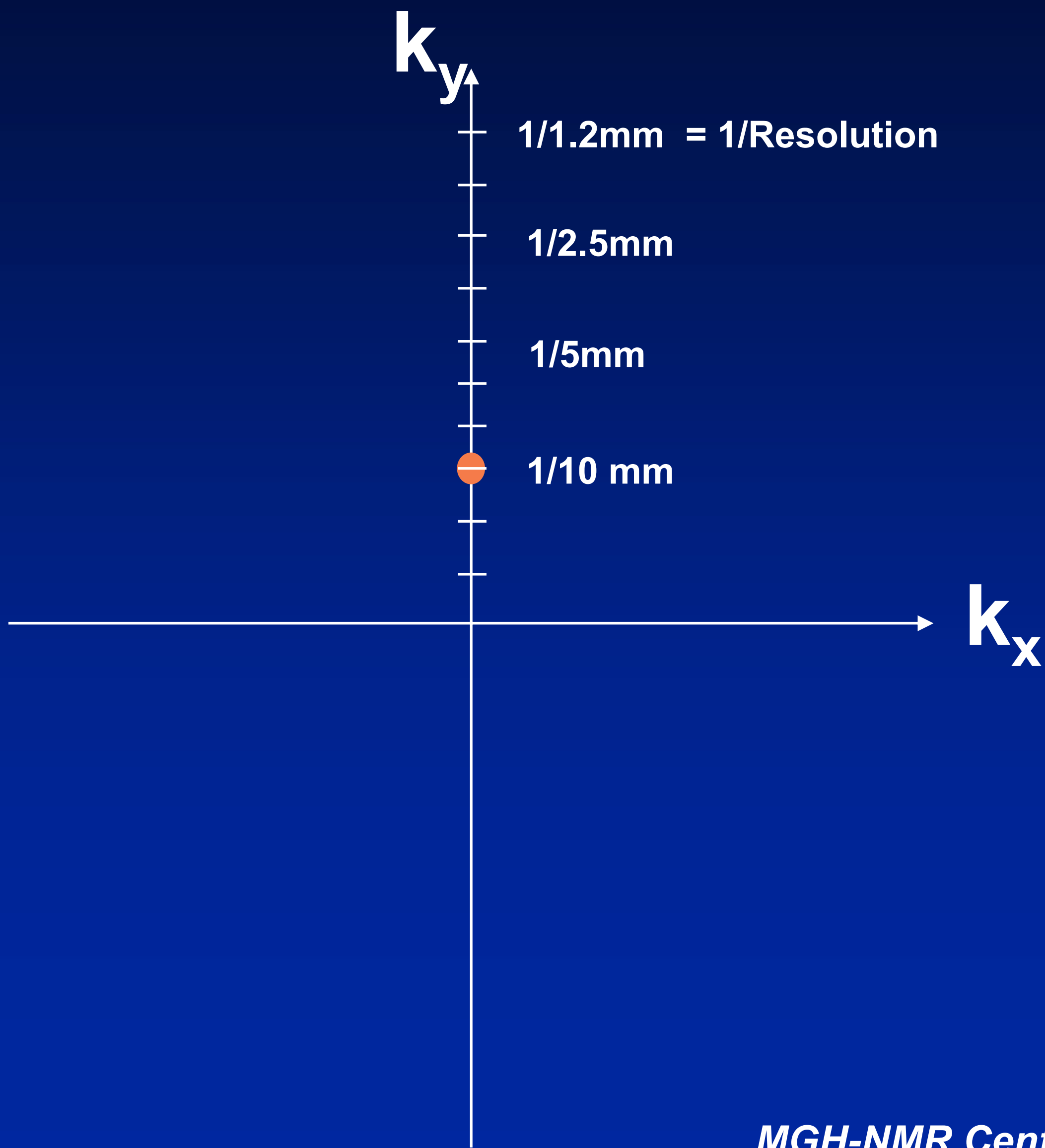
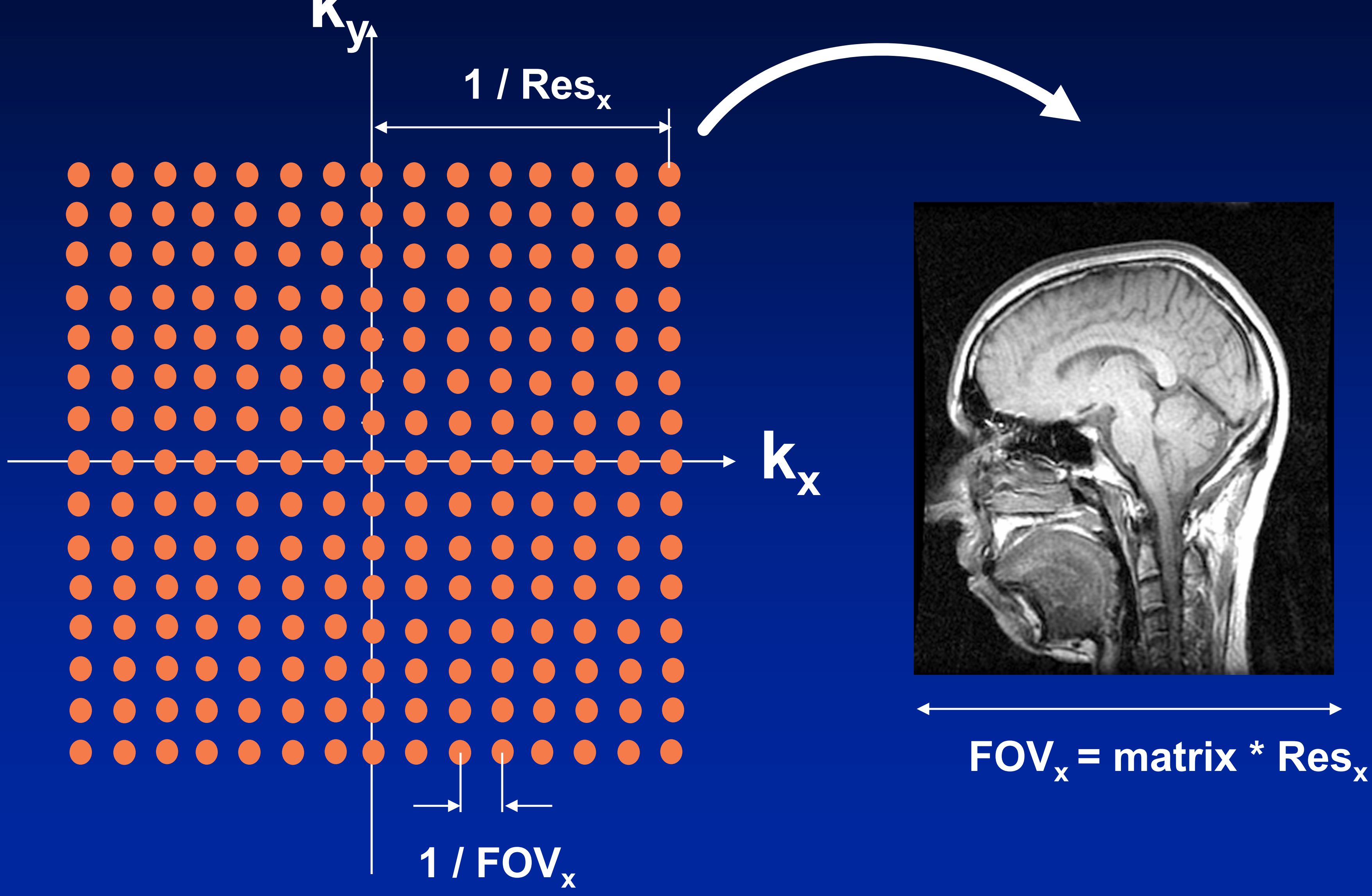
Measurement intensity at a spatial frequency...

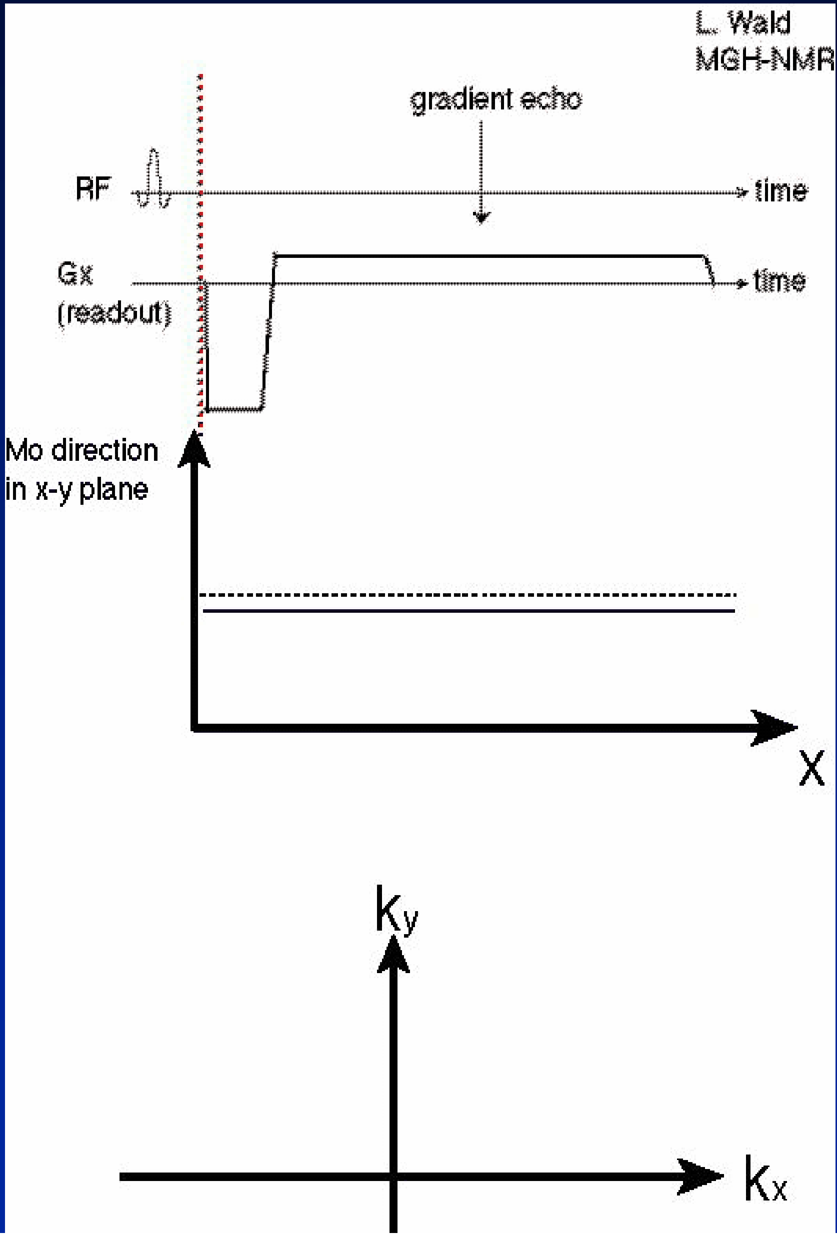


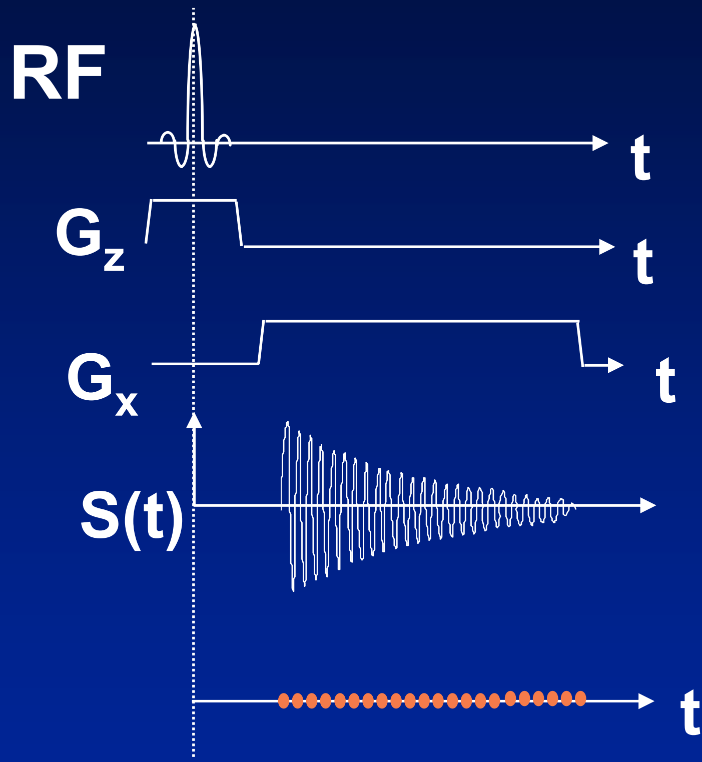


Fourier transform

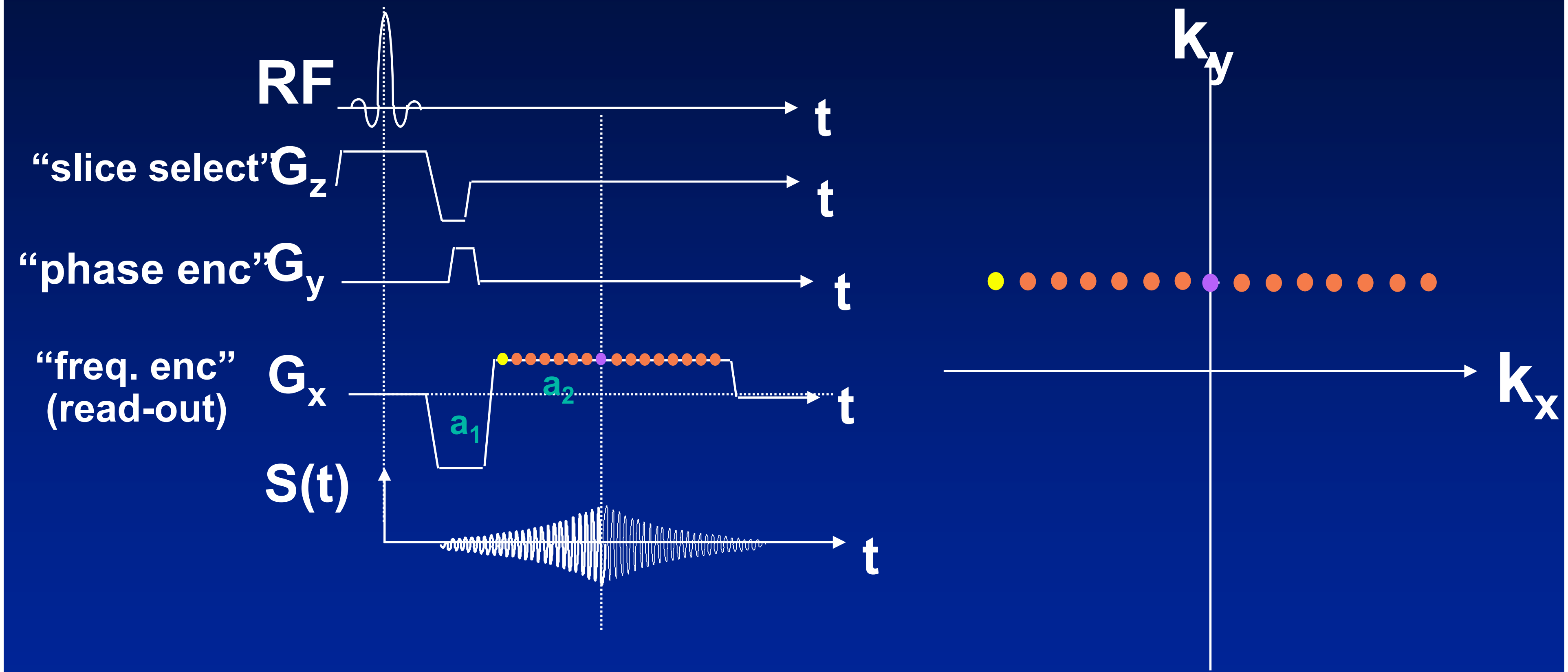


Frequency encoding revisited





“Spin-warp” encoding



one excitation, one line of kspace...

"Spin-warp" encoding mathematics

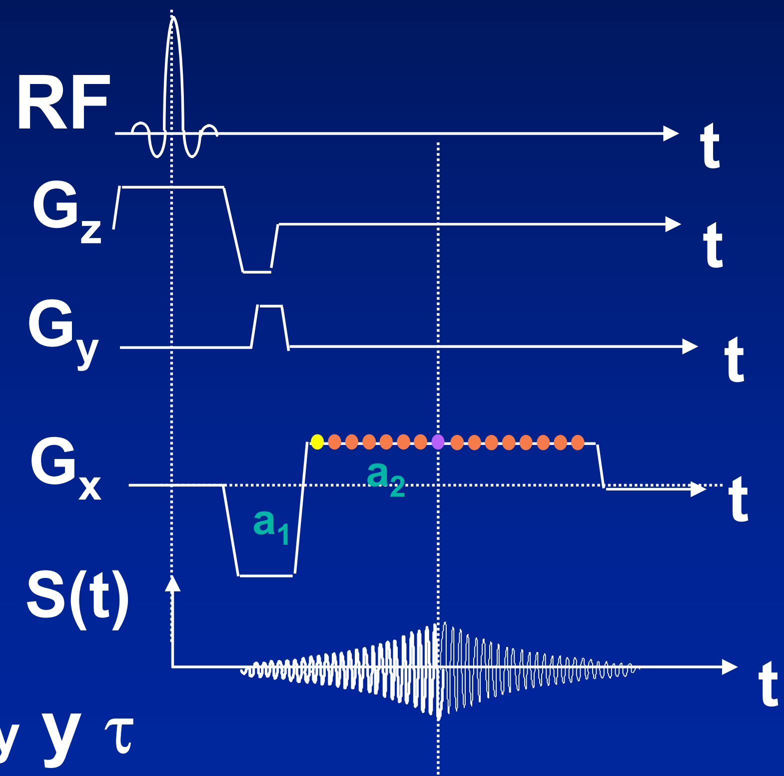
The "image" is the spin density function: $\rho(x)$

Phase due to readout:

$$\theta(t) = \omega_o t + \gamma G_x x t$$

Phase due to P.E.

$$\theta(t) = \omega_o t + \gamma G_y y \tau$$



$$\Delta\theta(t) = \omega_o t + \gamma G_x x t + \gamma G_y y \tau$$

"Spin-warp" encoding mathematics

Signal at time t from location (x,y)

$$S(t) \quad \rho(x,y)e^{i\gamma G_x xt + i\gamma G_y y\tau}$$

The coil integrates over object:

$$S(t) \quad \iint_{object} \rho(x,y)e^{i\gamma G_x xt + i\gamma G_y y\tau}\,dxdy$$

Substituting $k_x = -\gamma\ G_x\ t$ and $k_x = -\gamma\ G_x\ t$:

$$S(k_x,k_y) \quad \iint_{object} \rho(x,y)e^{-ik_x x - ik_y y}\,dxdy$$

"Spin-warp" encoding mathematics

View signal as a matrix in kx, ky…

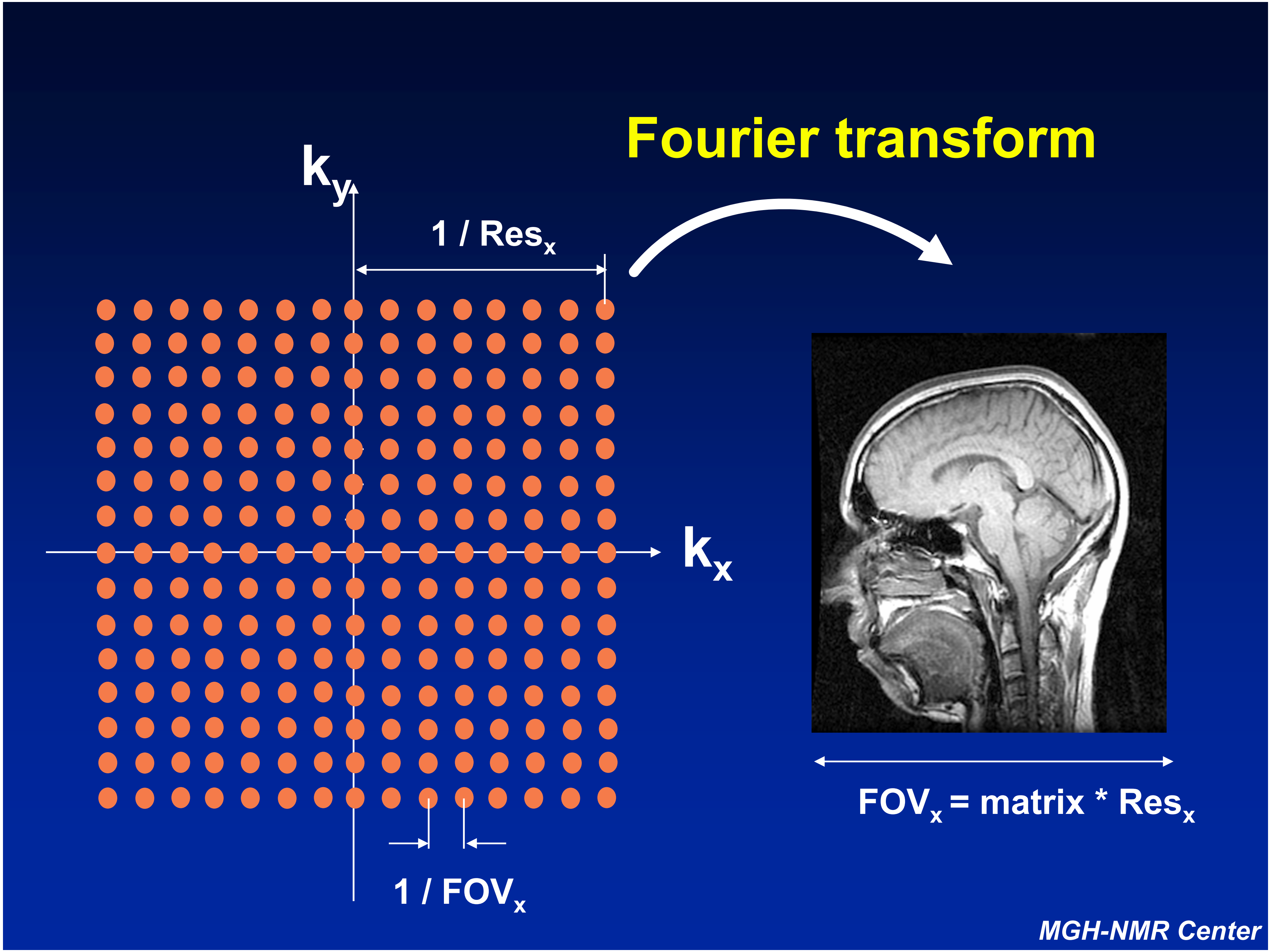
$$S(k_x, k_y) \quad \iint_{object} \rho(x,y) e^{-ik_x x - ik_y y} dxdy$$

:

Solve for ρ(x,y,)

$$\rho(x,y) \quad FT^{-1}\left[S(k_x, k_y)\right]$$

$$\rho(x,y) \quad \iint_{kspace} S(k_x, k_y) e^{ik_x x + ik_y y} dk_x dk_y$$



Fourier transform
k_y
$1 / Res_x$
k_x
$1 / FOV_x$
$FOV_x = matrix * Res_x$
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Kspace facts

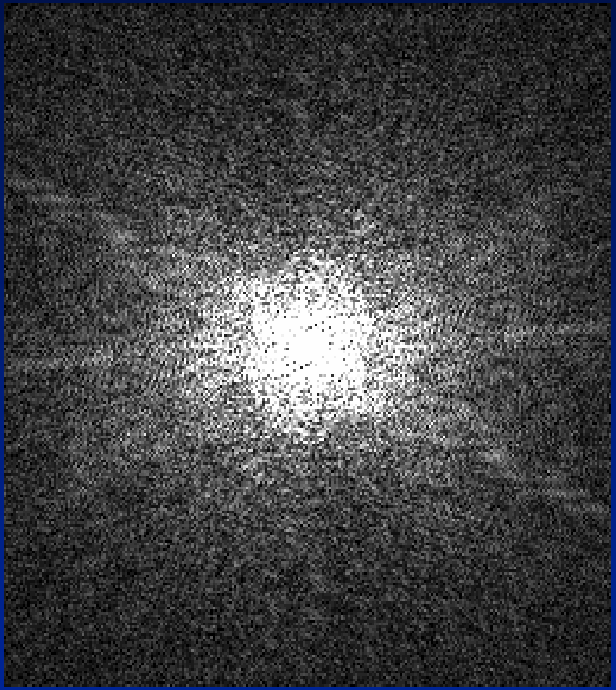
Resolution is determined by the largest spatial freq sampled.

FOV = matrix * resolution

If the object is real, half the information in kspace matrix is redundant. We only need to record half of it.

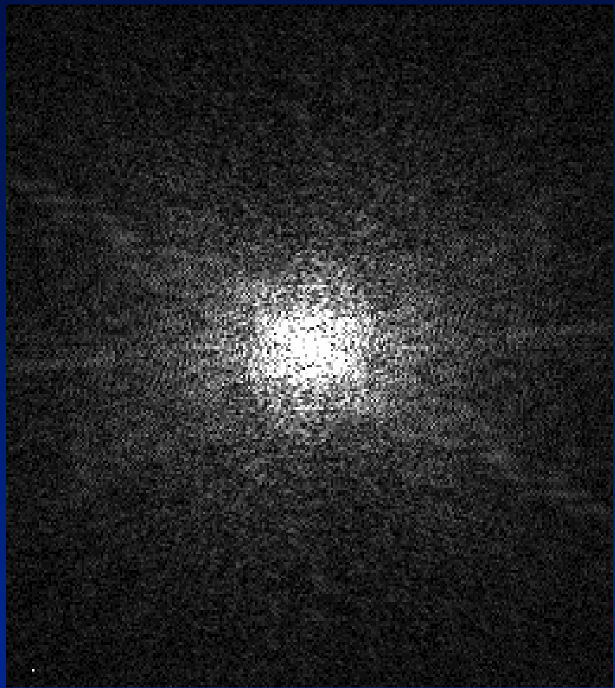
kspace

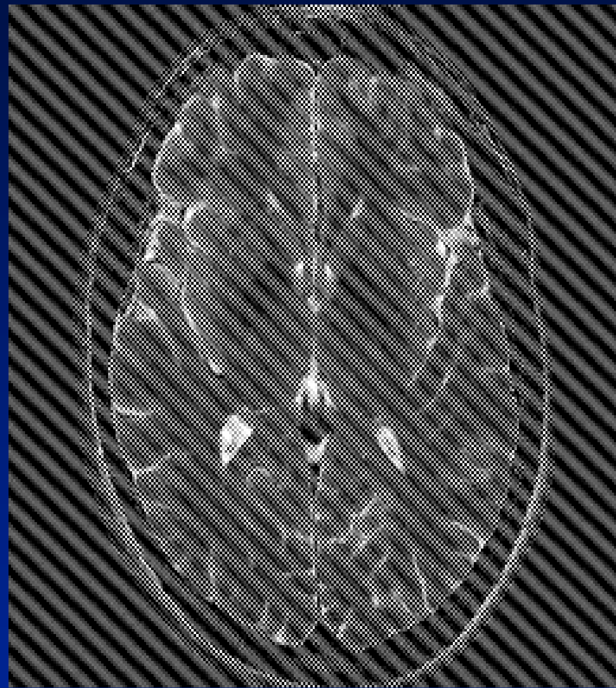
Image space (magnitude)



kspace (magnitude)

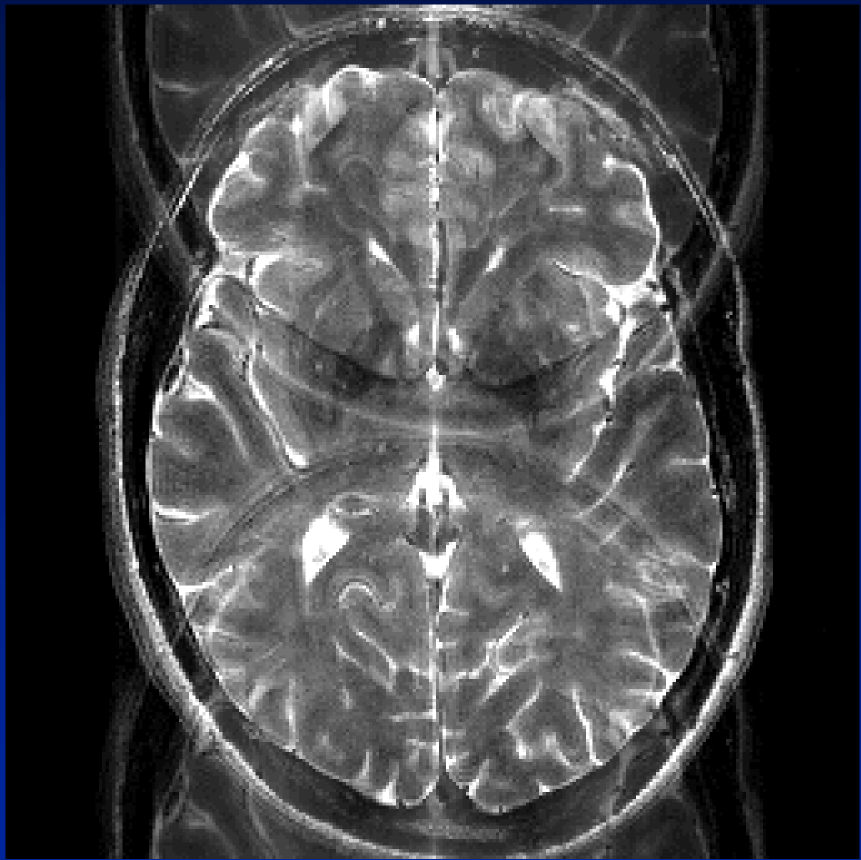
kspace artifacts: spike





One “white pixel” in kspace from a electric spark

Kspace artifacts: Symmetric N/2 ghost



Even numbered lines got $\exp(i\phi)$

Odd numbered lines got $\exp(-i\phi)$

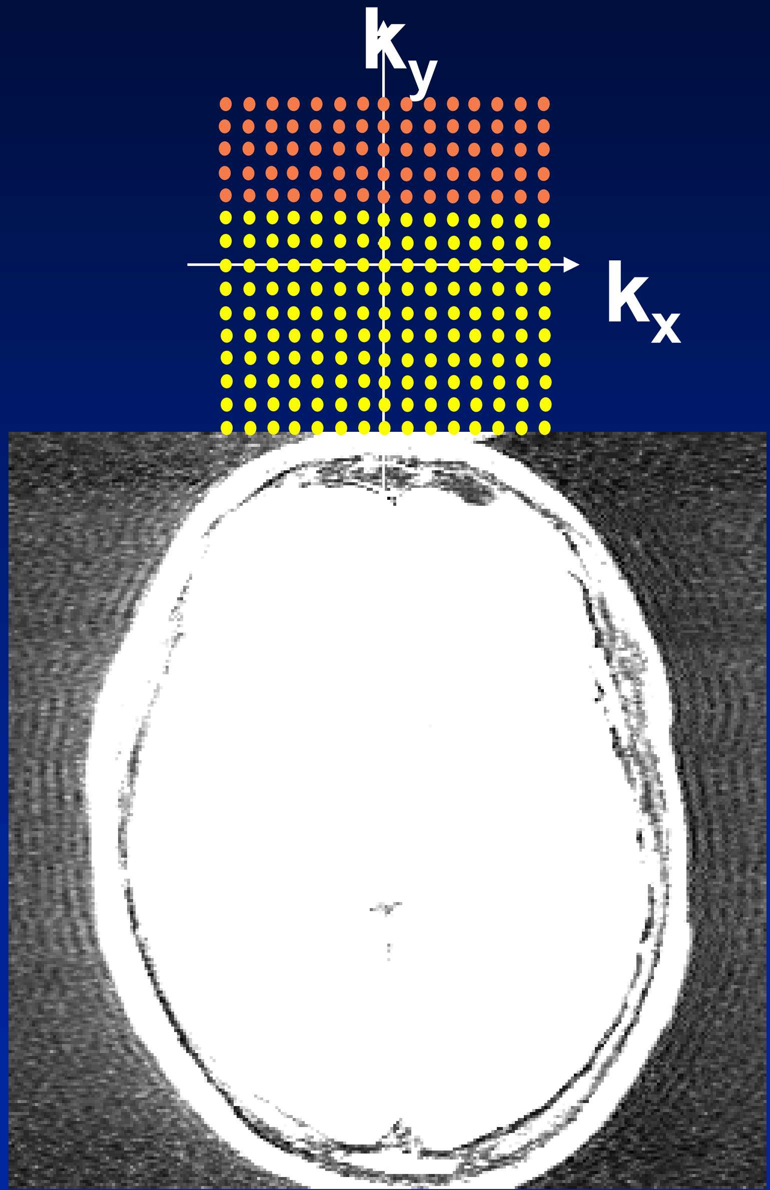
ϕ = 12 degrees

kspace artifacts: subject motion

Yellow = position1
Orange = moved 2 pixels

Movement in real space = linear phase shift across kspace.

=> Orange points have linear phase $\theta = a\, k_y$



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T1-Weighting

Very long TR, Signal ~ Proton Density

Shortening TR $\rightarrow$ long T1 darker

"Best" T1 contrast, TR near T1

Contrast comes at expense of Signal: throw away some magnetization

Fast Imaging

"Dost thou love life?
Then do not squander time,
for that's the stuff
life is made of."

- Benjamin Franklin

Requirements for brain mapping

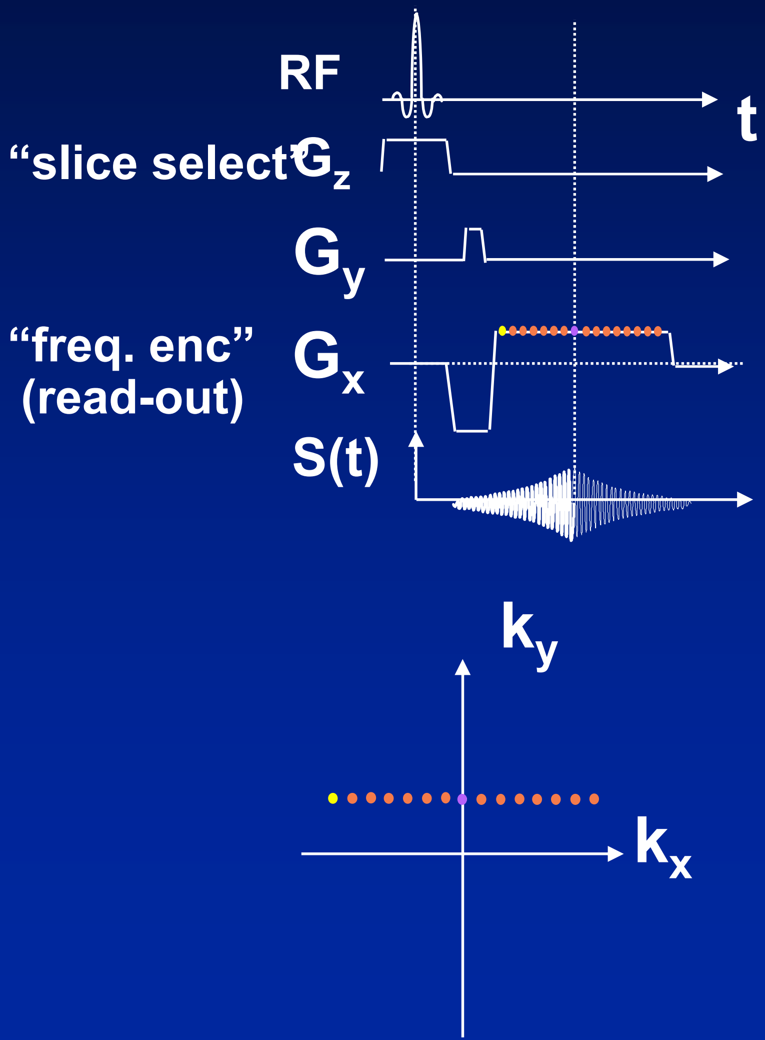
Considerations:

- **Signal increase = 0 to 5% (small)**
- **Motion artifact on conventional image is 0.5% - 3%**
- **Need to see changes on timescale of hemodynamic changes (seconds)**

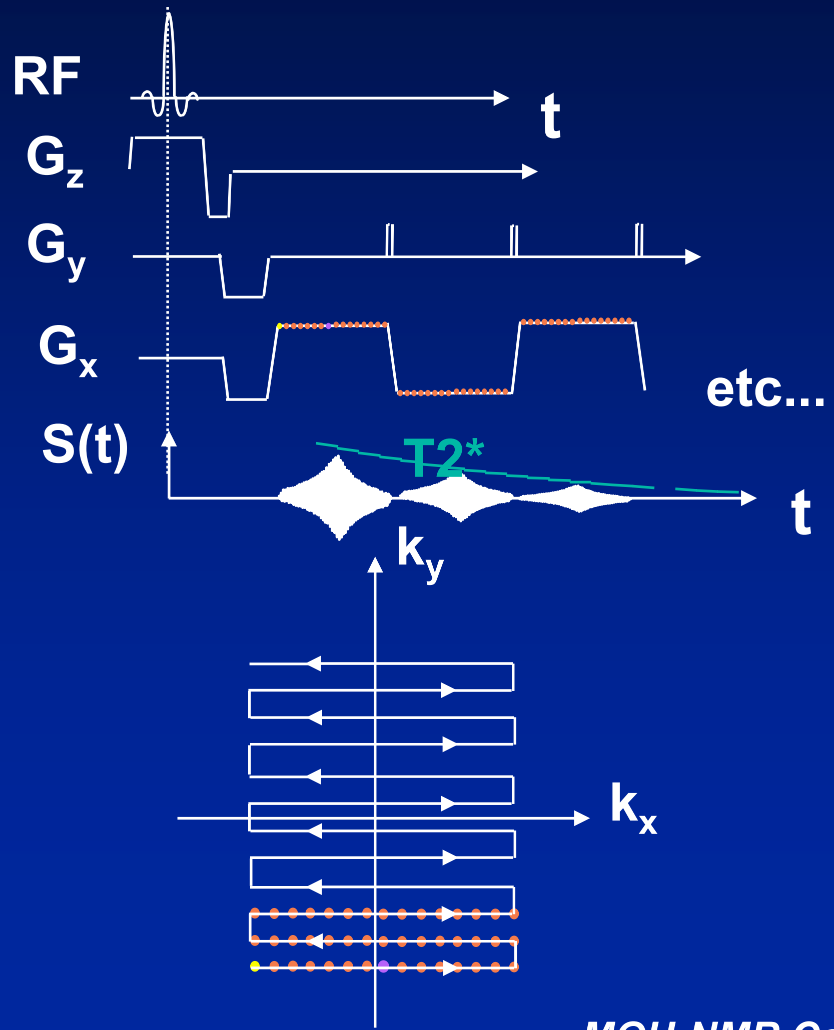
Requirement: Fast, "single shot" imaging, image in 80ms, set of slices every 1-3 seconds.

What's the difference?

conventional MRI



echoplanar imaging



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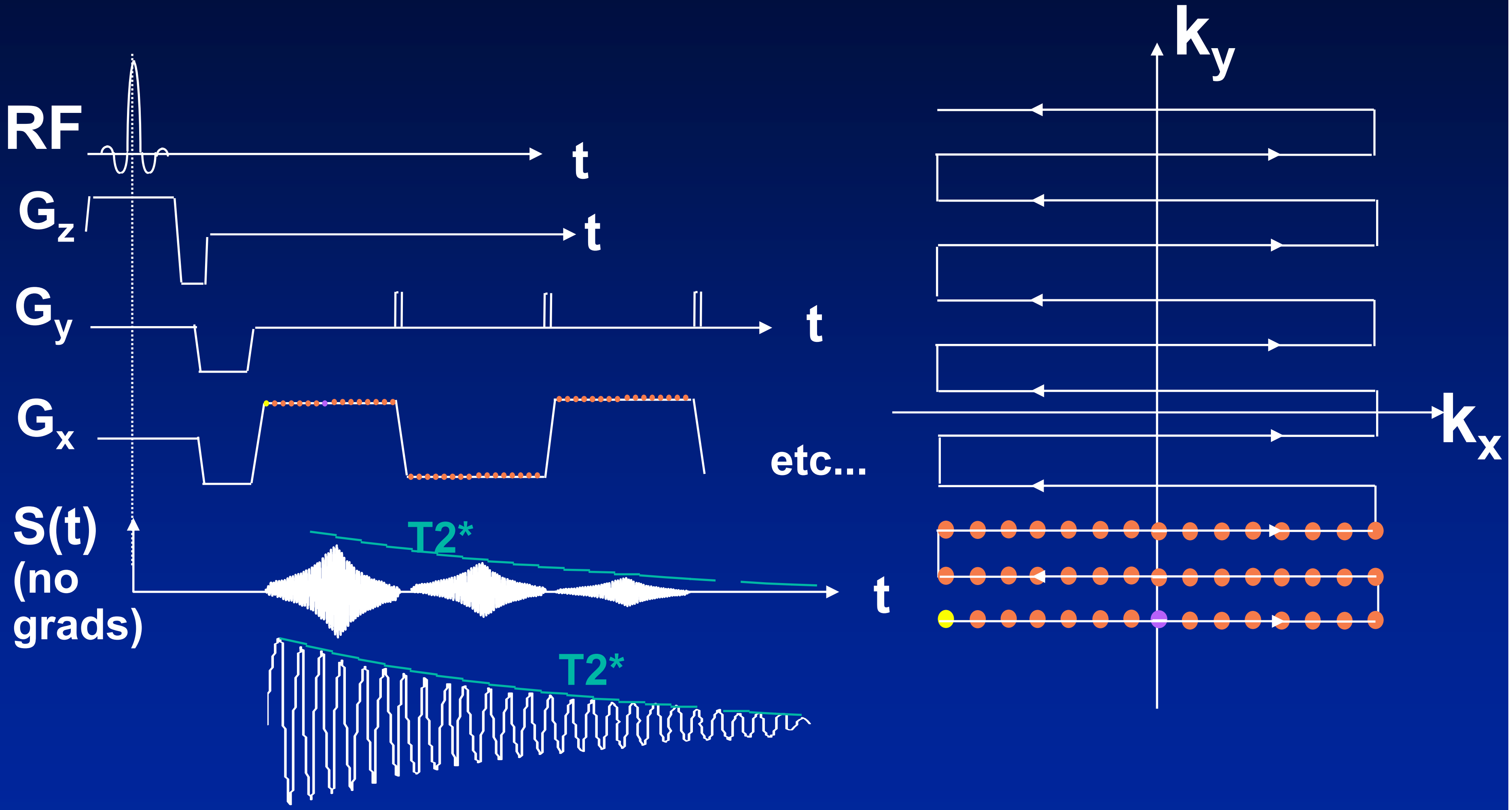
"Echo-planar" encoding



one excitation, many lines of kspace...

“Echo-planar” encoding

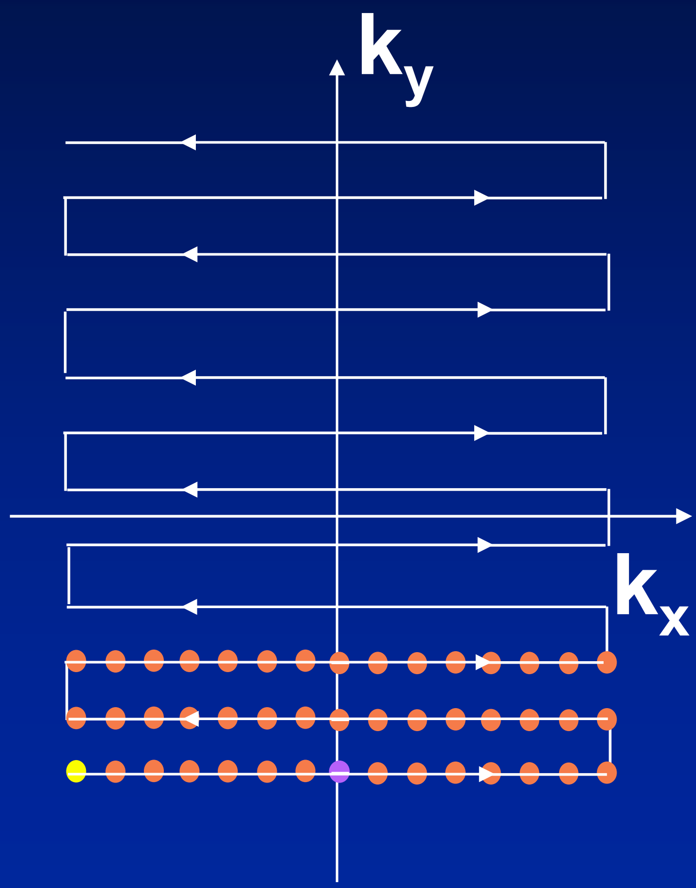
Observations:

- Adjacent points along kx are taken with short Δt (= 5 us). (high bandwidth)

- Adjacent points along ky are taken with long Δt (= 500us). (low bandwidth)

- A given line is read quickly, but the total encode time is longer than conventional Imaging.

- Adjacent lines are traversed in opposite directions.

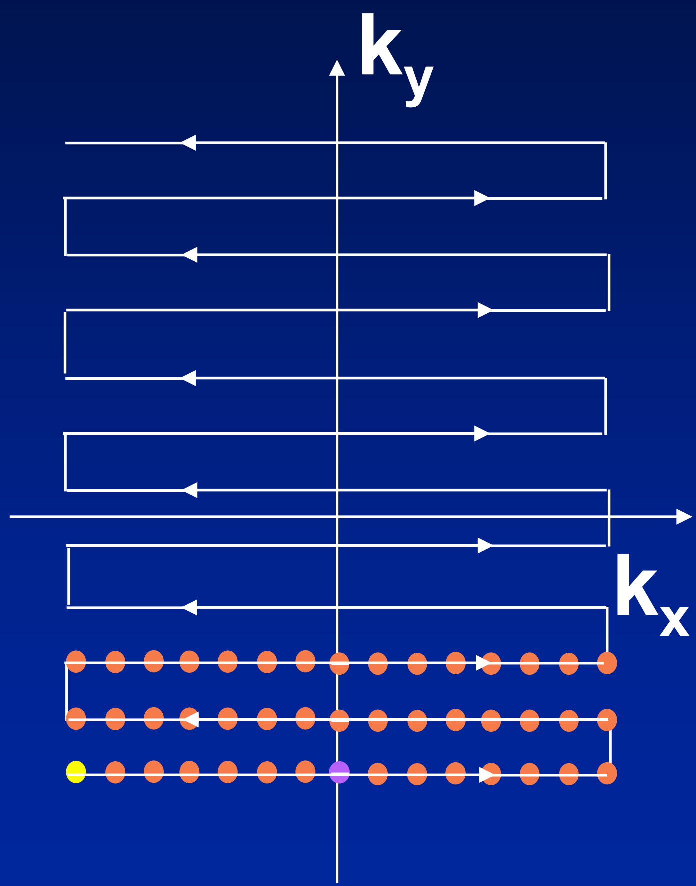


Bandwidth is asymmetric in EPI

• Adjacent points in k_x have short
  Δt = 5 us (high bandwidth)

• Adjacent points along ky are taken
with long Δt (= 500us). (low bandwidth)

The phase error (and thus distortions) are in the phase encode direction.

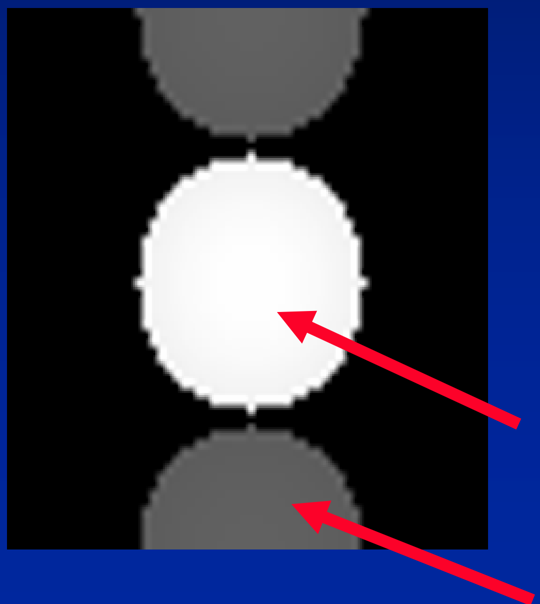


EPI problems: N/2 ghost

Asymmetry in alternate lines gives N/2 image ghost.

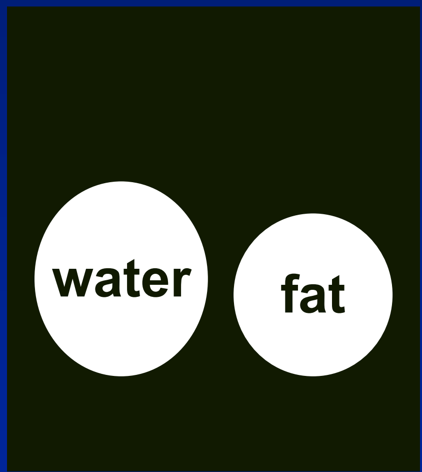
Asymmetry from:

Eddy currents
receiver filter
receiver timing
head coil tuning.

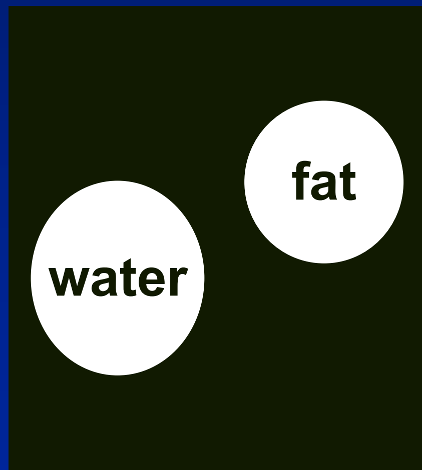


EPI problems: frequency offset

If one object has a different NMR frequency (e.g. fat and water) it gets shifted in PE direction. (why?)

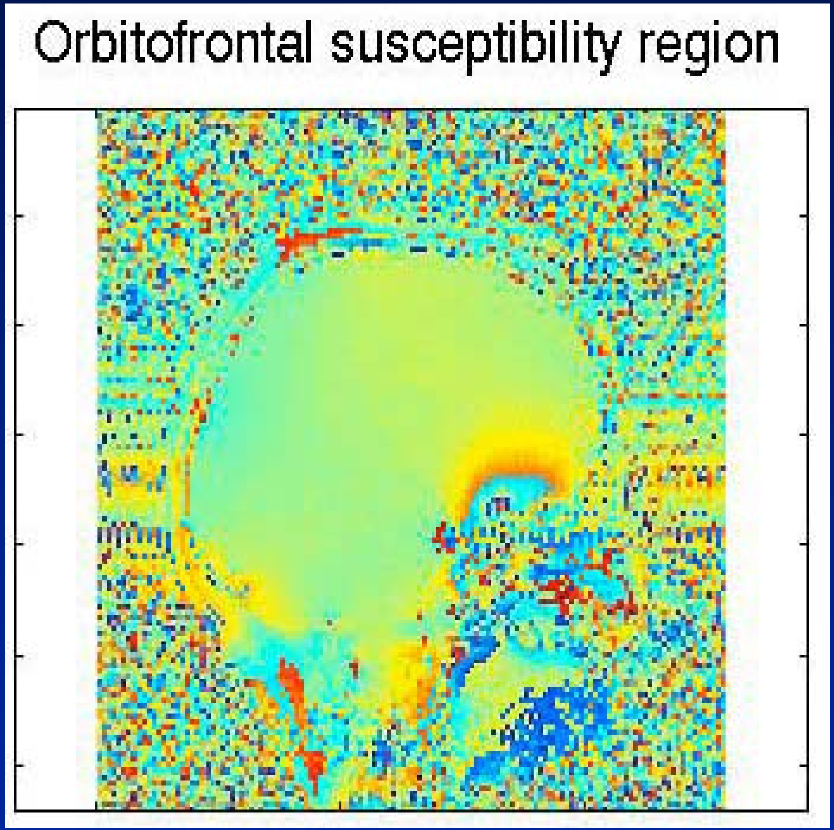


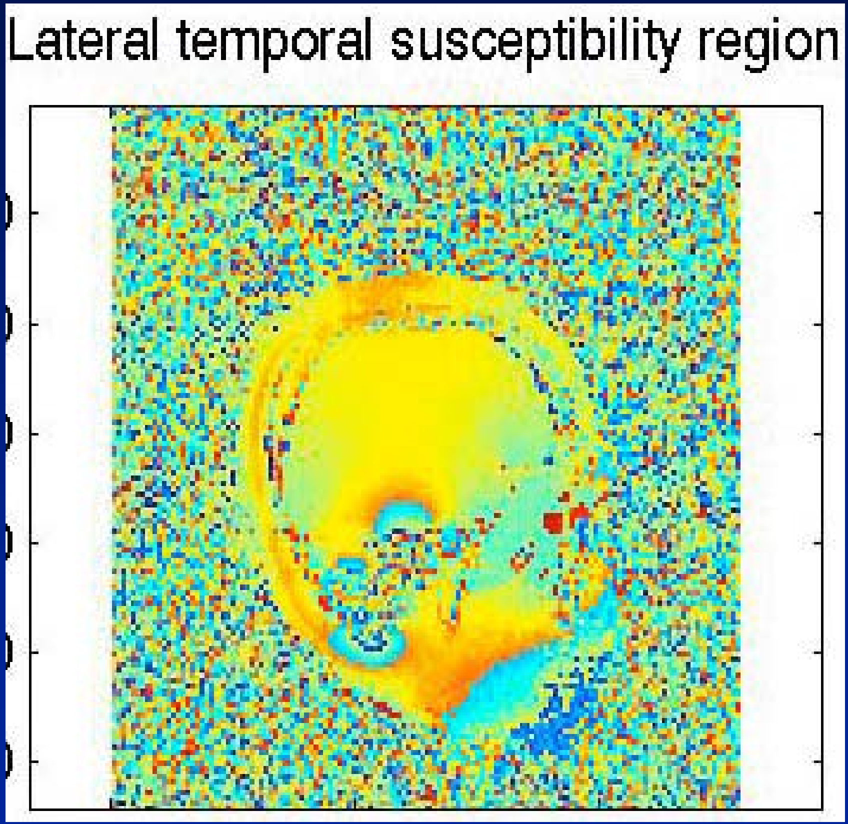
True location



Echoplanar image

Enemy #1 of EPI: local susceptibility gradients





B_0 field maps in the head

What do we mean by "susceptibility"?

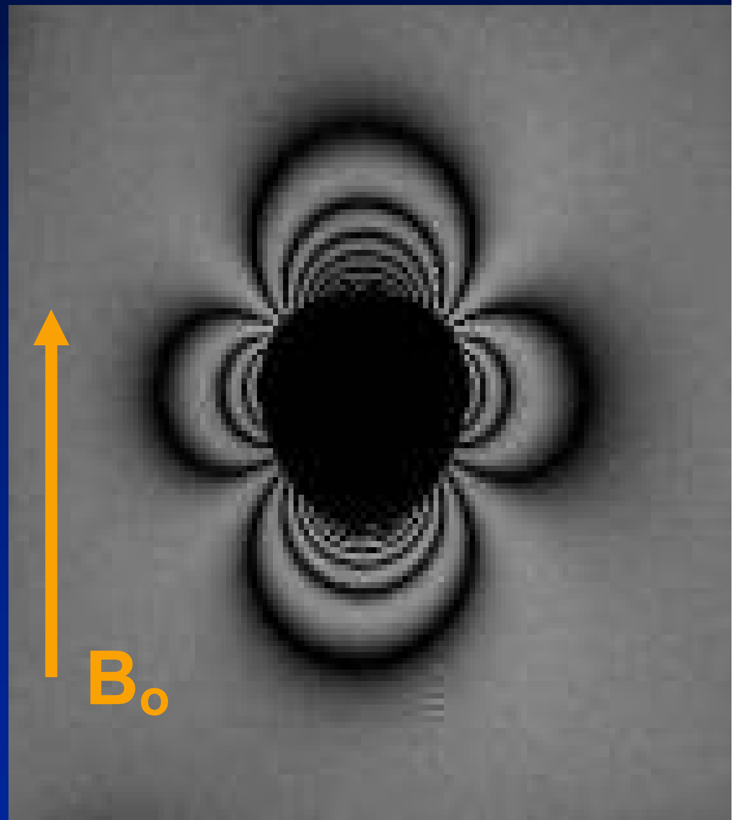
In physics, it refers to a material's tendency to magnetize when placed in an external field.

In MR, it refers to the effects of magnetized material on the image through its local distortion of the static magnetic field B_o.

Ping-pong ball in water…

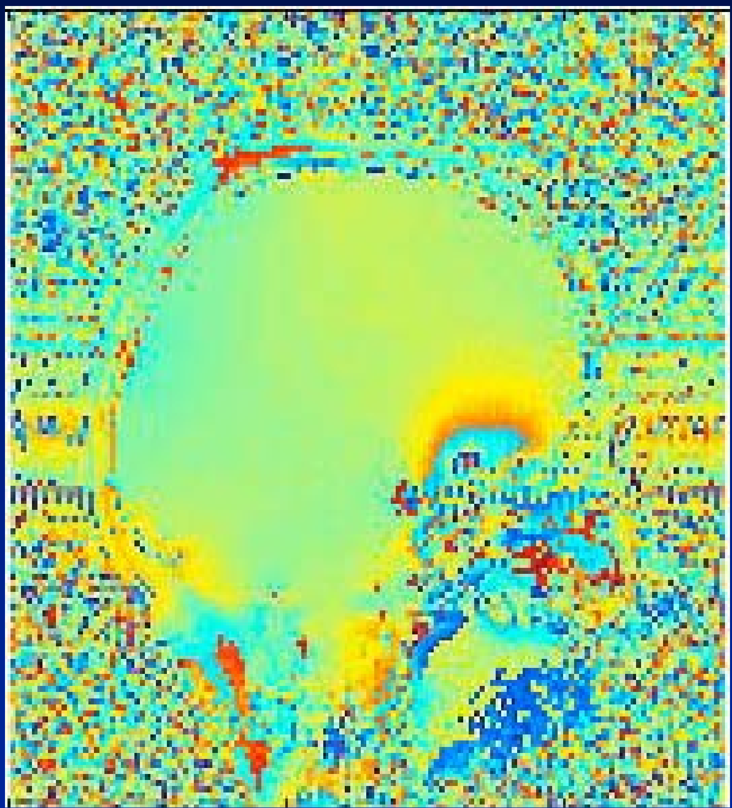
Susceptibility effects occur near magnetically dis-similar materials

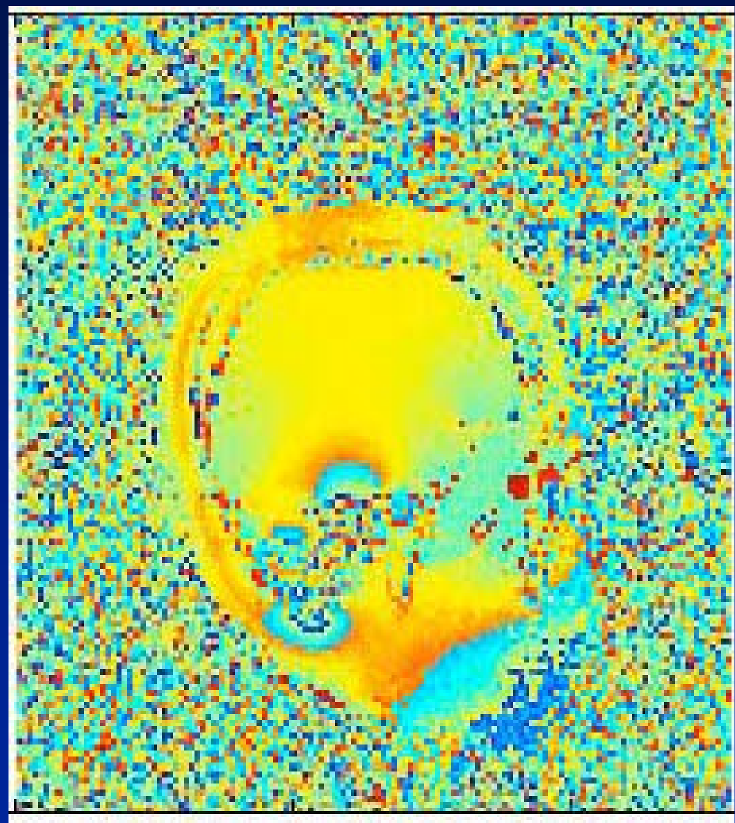
Field disturbance around air surrounded by water (e.g. sinuses)



Field map (coronal image) 1.5T

B_o map in head: it's the air tissue interface…

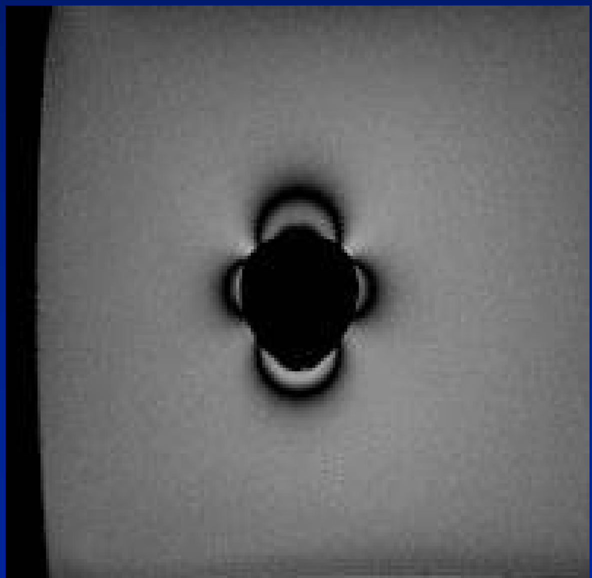




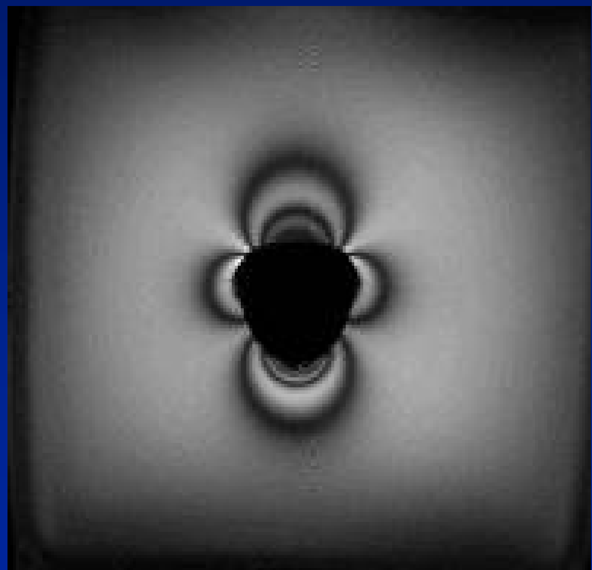
Sagittal Bo field maps at 3T

Susceptibility field (in Gauss) increases w/ B_o

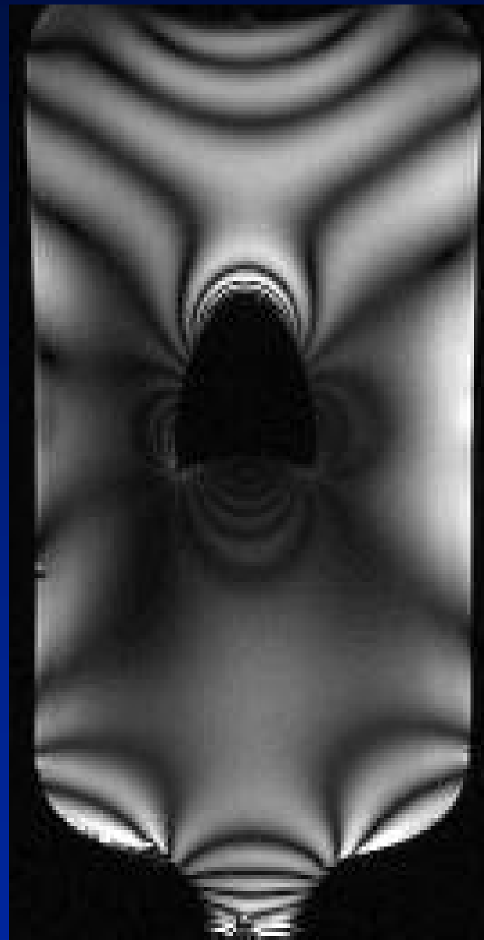
Ping-pong ball in H_20:
Field maps (ΔTE = 5ms), black lines spaced by 0.024G (0.8ppm at 3T)



1.5T



3T



7T

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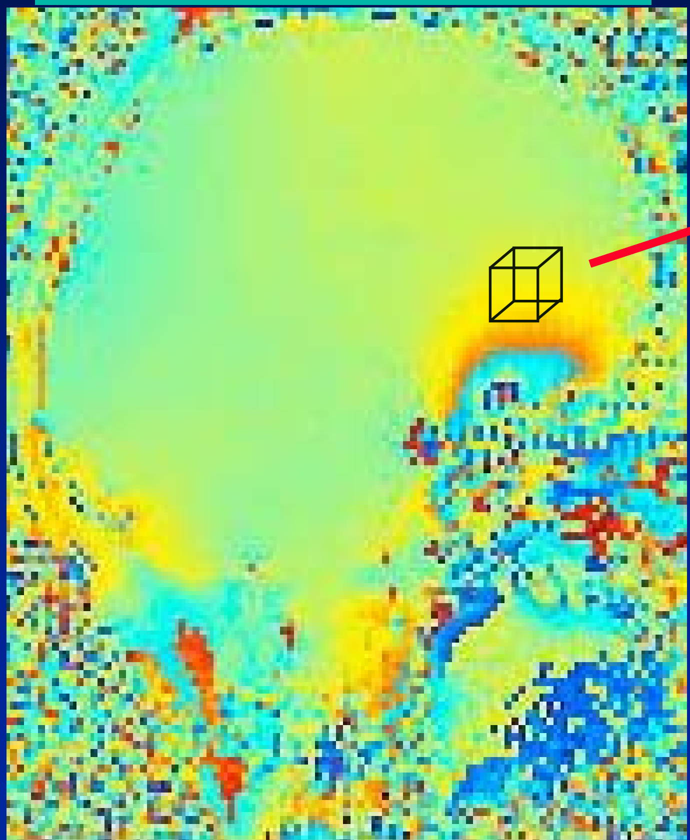
Local susceptibility gradients: 2 effects

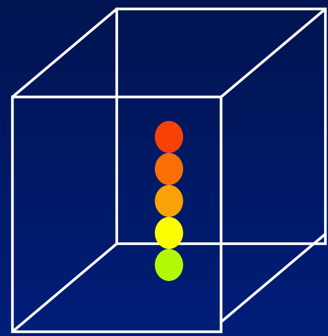
1) **Local dephasing of the signal (signal loss) within a voxel, mainly from <u>thru-plane</u> gradients**

2) **Local geometric distortions, (voxel location improperly reconstructed) mainly from local <u>in-plane</u> gradients.**

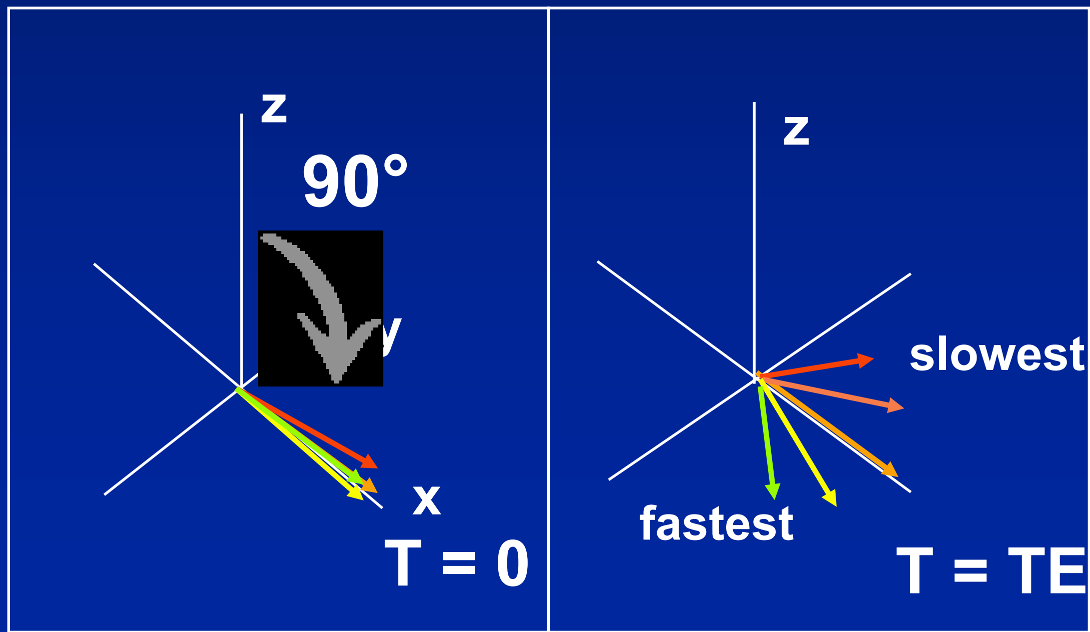
1) Non-uniform Local Field Causes Local Dephasing

Sagittal B_o field map at 3T



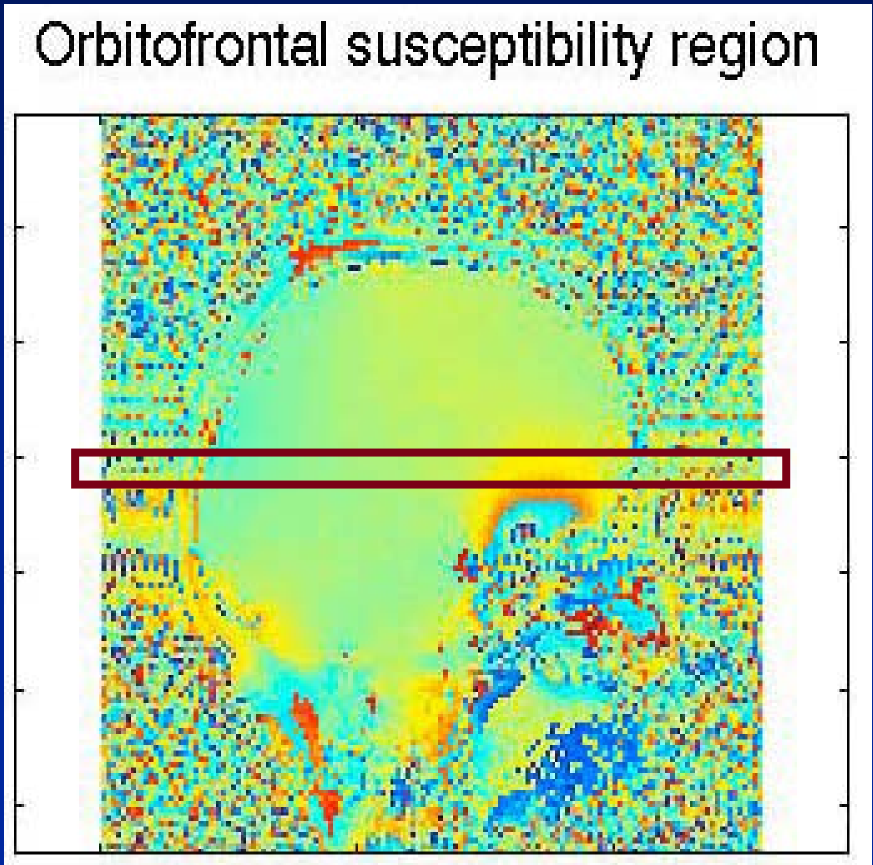


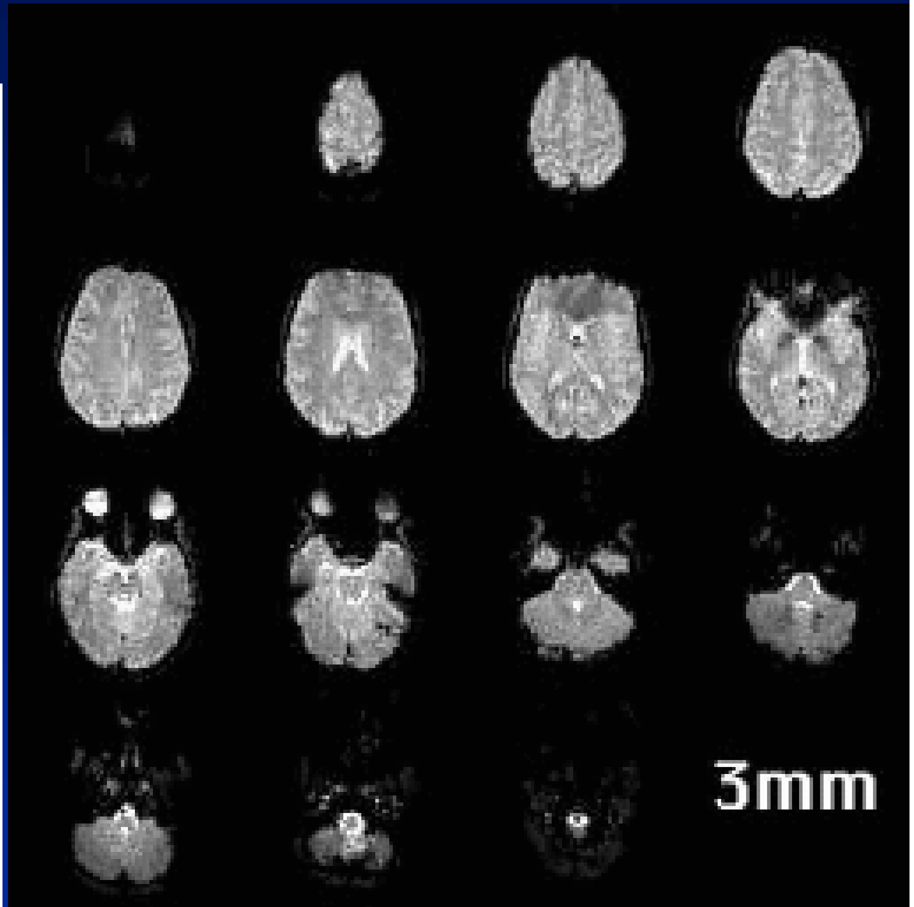
5 water protons in different parts of the voxel…



Local susceptibility gradients: thru-plane dephasing

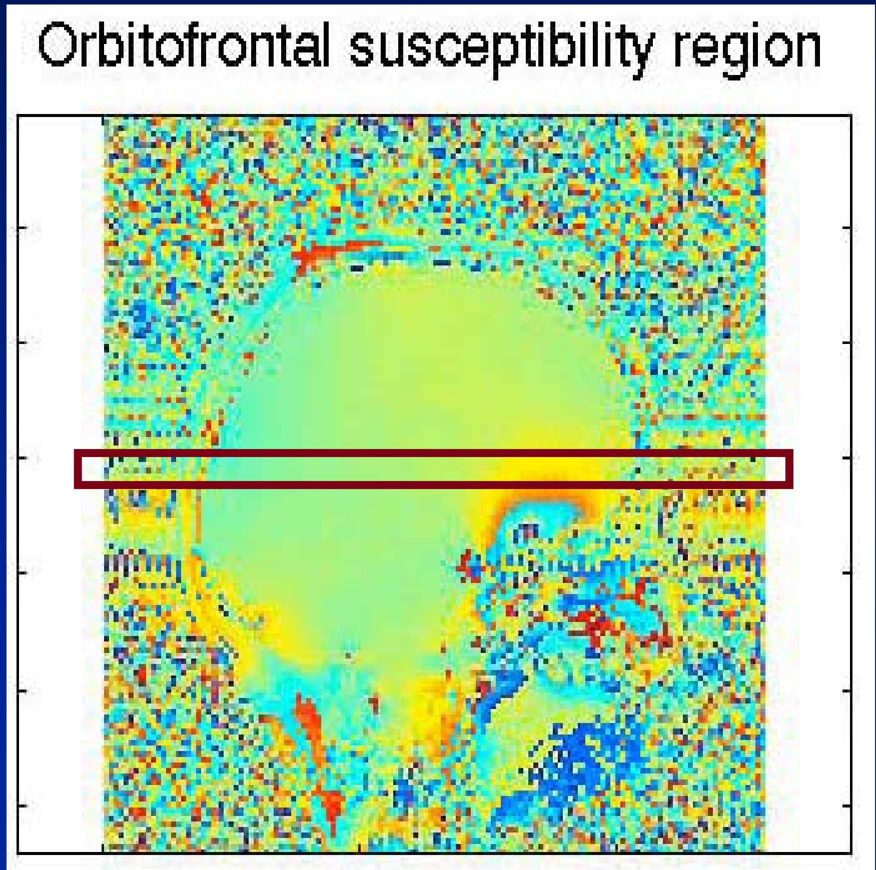
Bad for thick slice above frontal sinus…

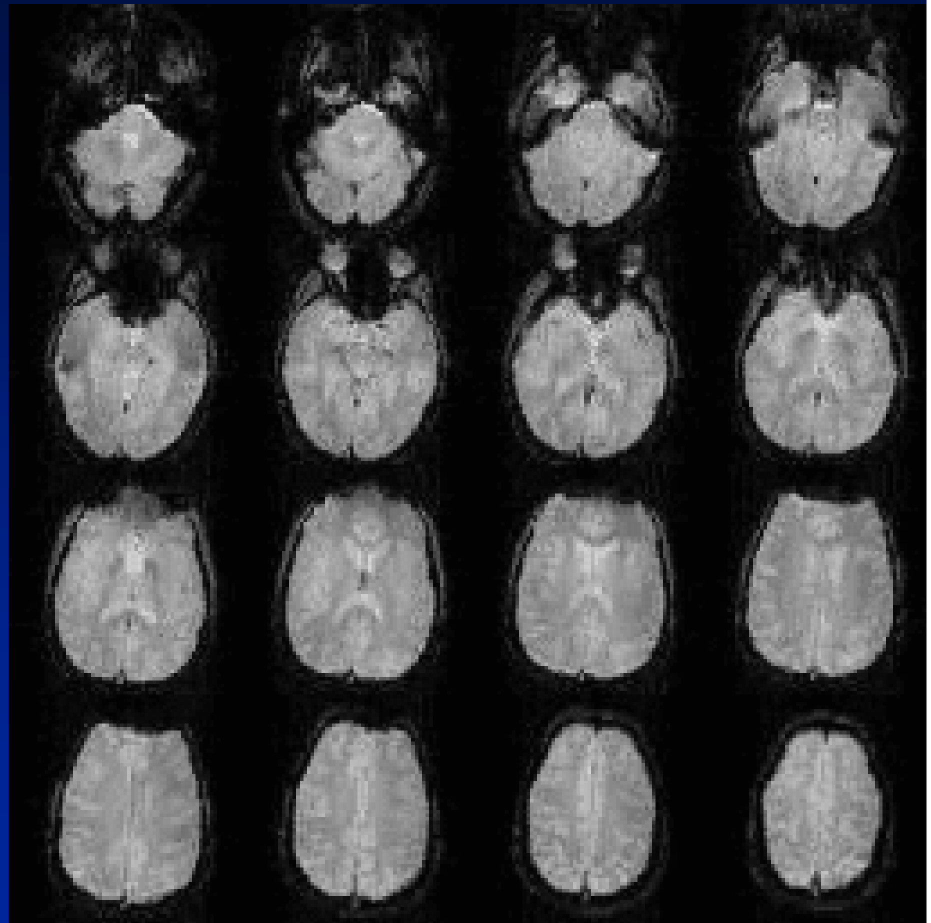




3T

Thru-plane dephasing gets worse at longer TE





3T, TE = 21, 30, 40, 50, 60ms

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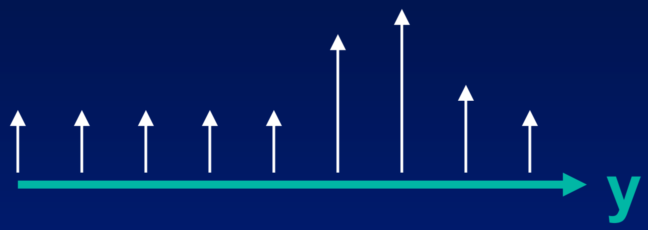
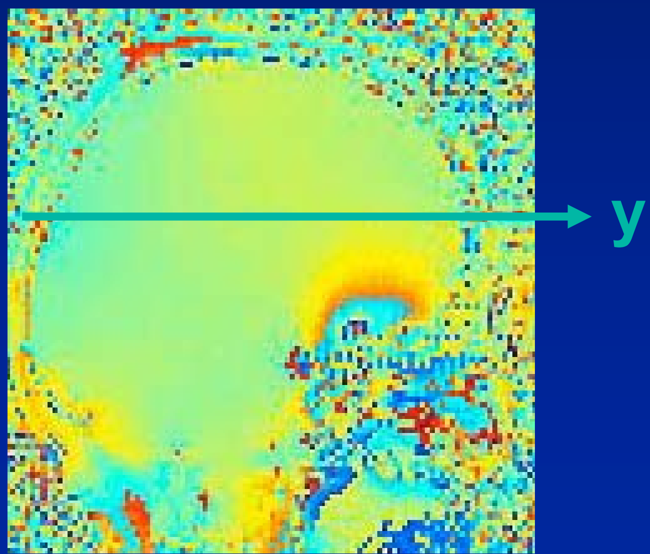
Problem #2 Susceptibility Causes Image Distortion in EPI



Field near sinus

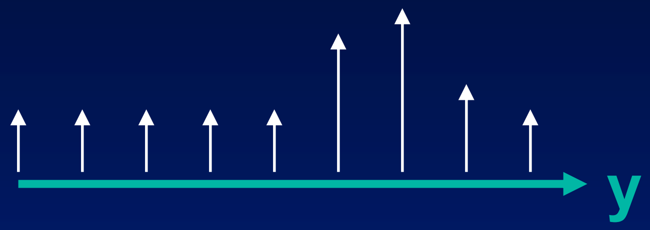


To encode the image, we control phase evolution as a function of position with applied gradients.

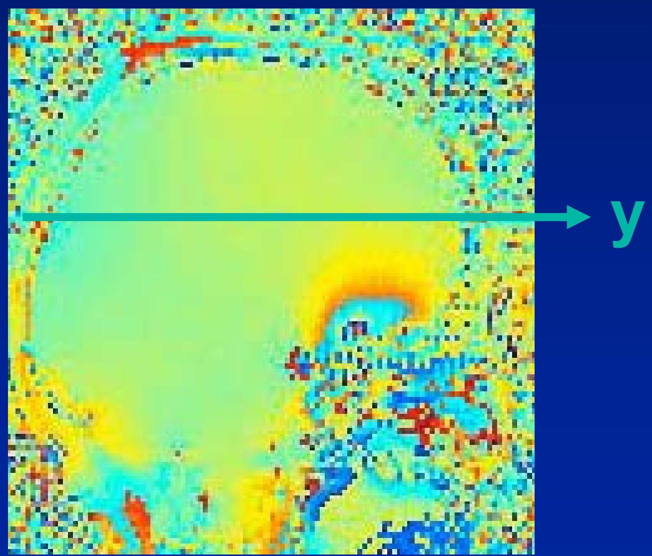
Local suscept. Gradient causes unwanted phase evolution.

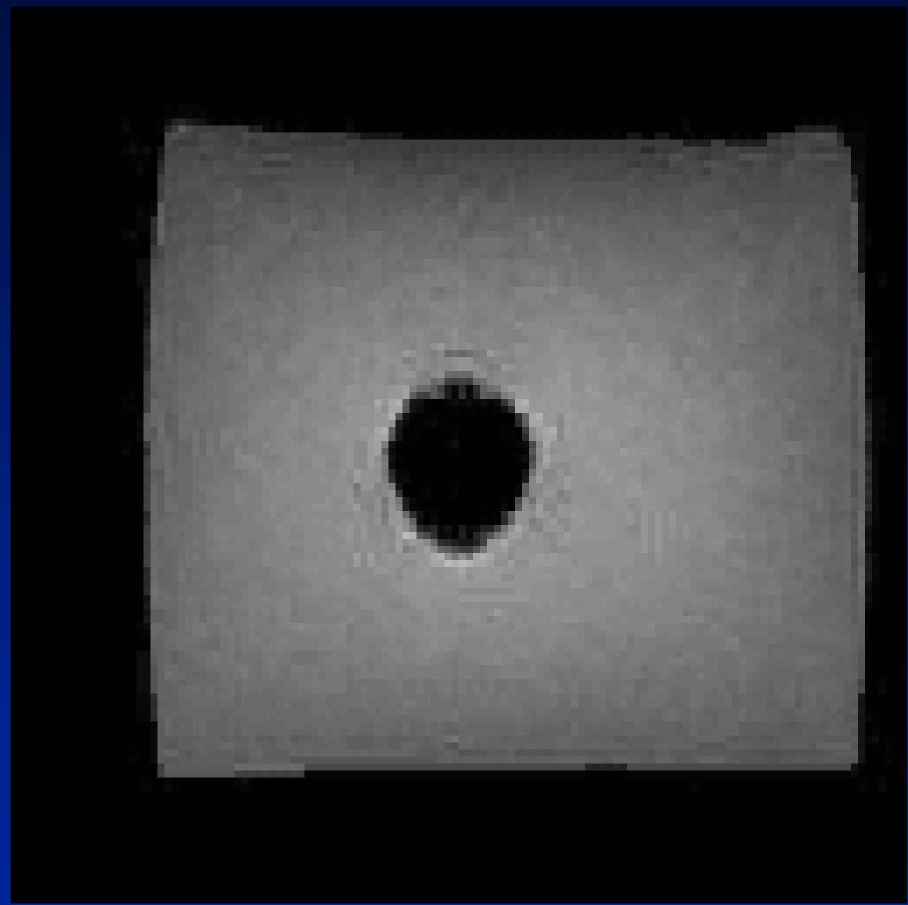
The phase encode error builds up with time. $\Delta\theta = \gamma B_{local} \Delta t$

Susceptibility Causes Image Distortion



Field near
sinus

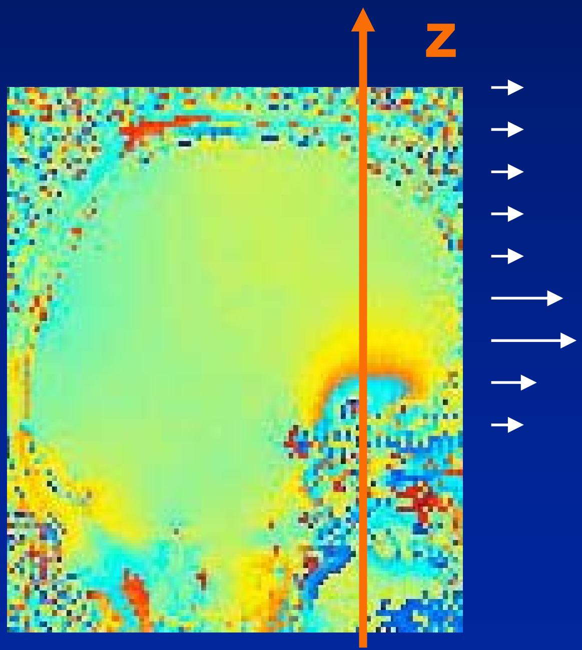




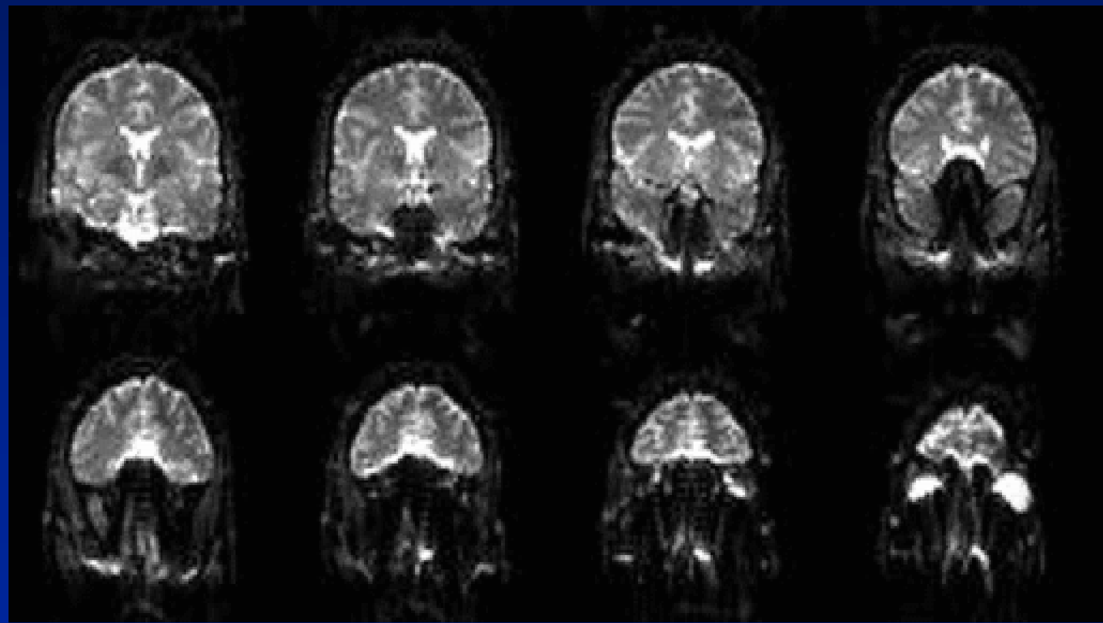
Conventional grad. echo,
$\Delta\theta \propto$ encode time $\propto$ 1/BW

Susceptibility Causes Image Distortion

Echoplanar Image,
$\Delta\theta \propto$ encode time $\propto$ 1/BW



Field near sinus



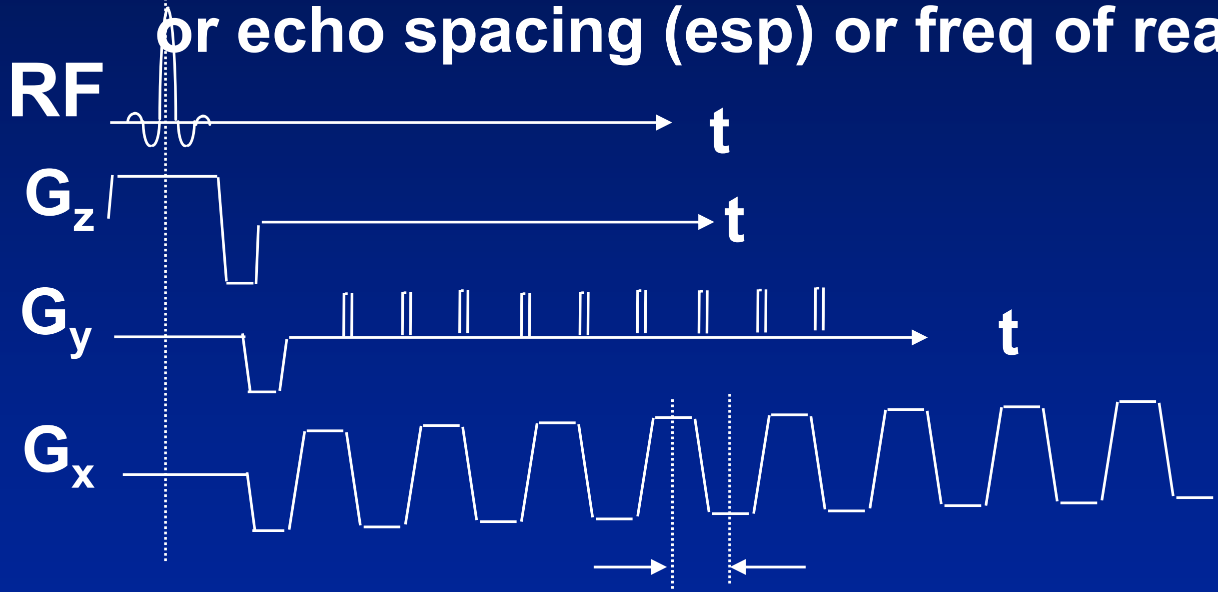
3T head gradients

Encode time = 34, 26, 22, 17ms

Characterization of grad. performance

- length of readout train for given resolution

or echo spacing (esp) or freq of readout…



'echo spacing' (esp)

esp = 500 us for whole body grads, readout length = 32 ms
esp = 270us for 3T, readout length = 17 ms

What is important in EPI performance?

Short image encoding time.

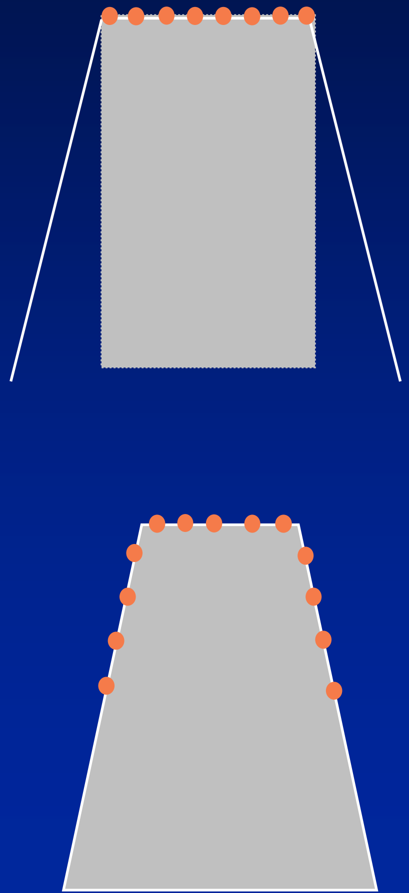
Parameters related to total encoding time:
1) echo spacing.
2) frequency of readout waveform.

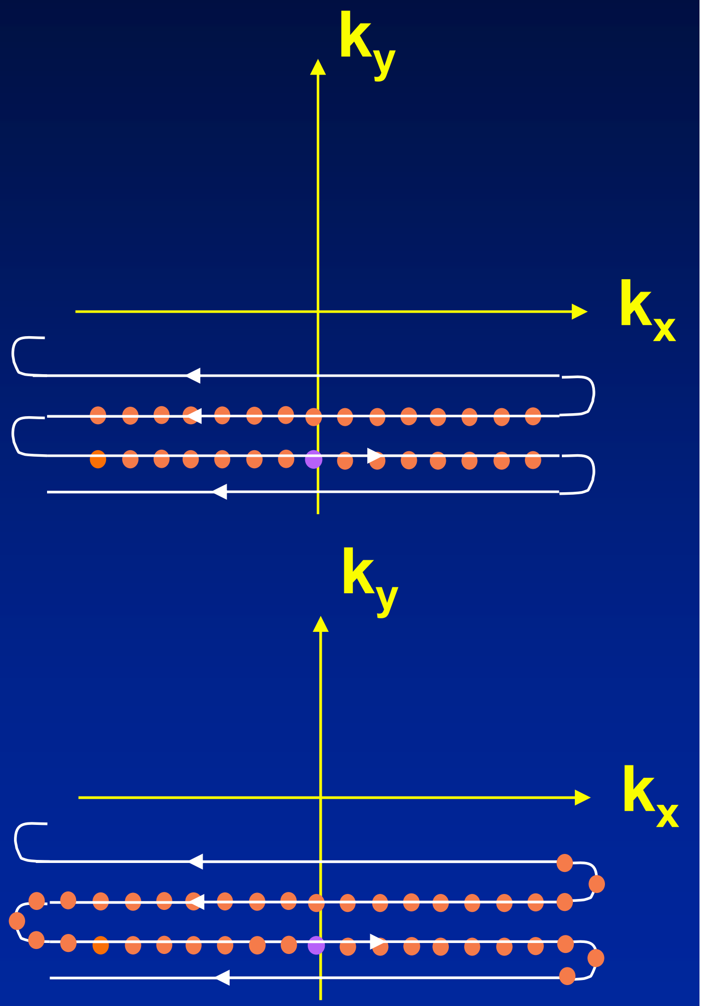
Key specs for achieving short encode times:
1) gradient slew rate.
2) gradient strength.
3) ability to ramp sample.

Good shimming (second order shims)

Ramp-sampling of EPI is key to reducing encode time

Area under samples
α 1 / Resolution





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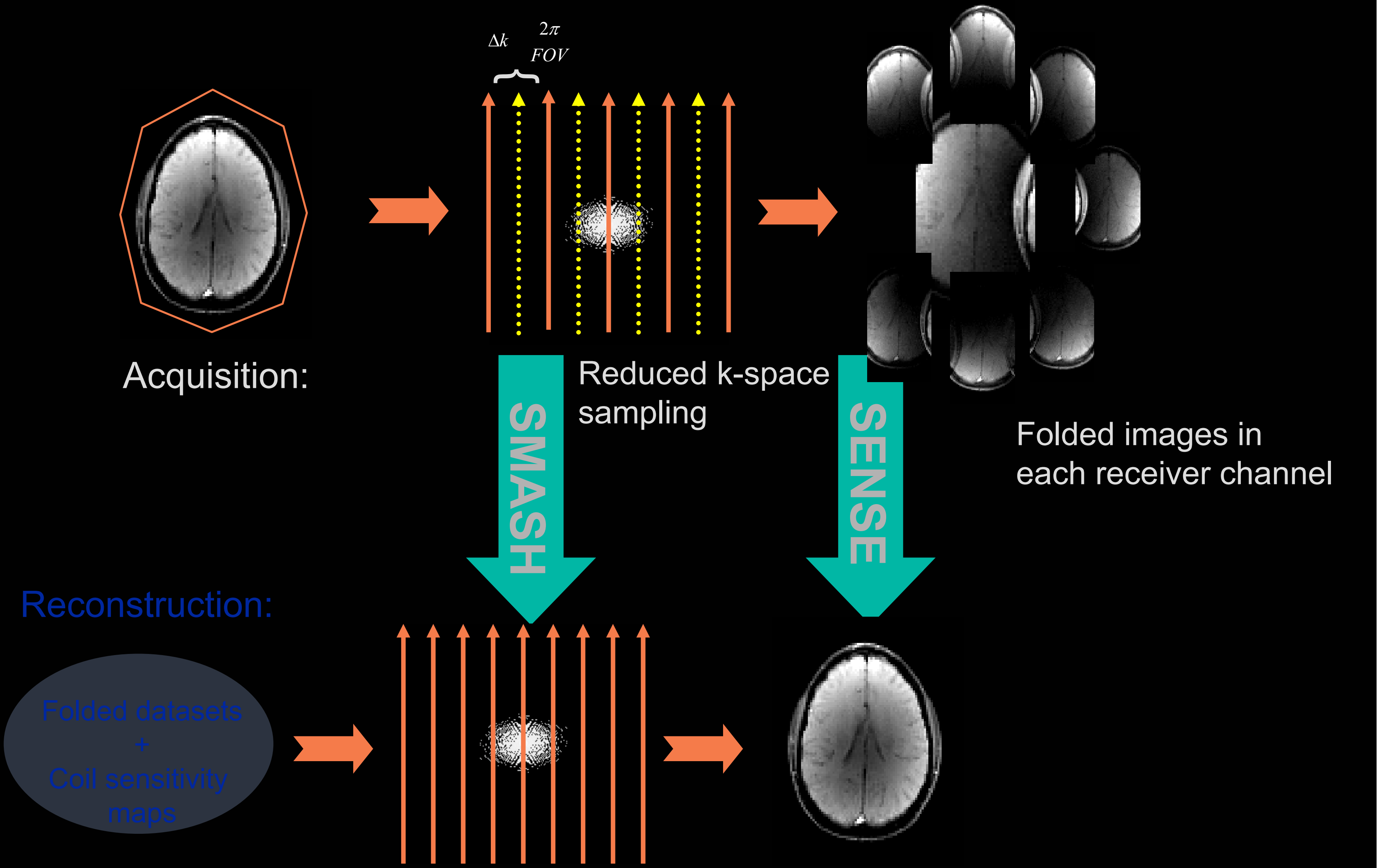
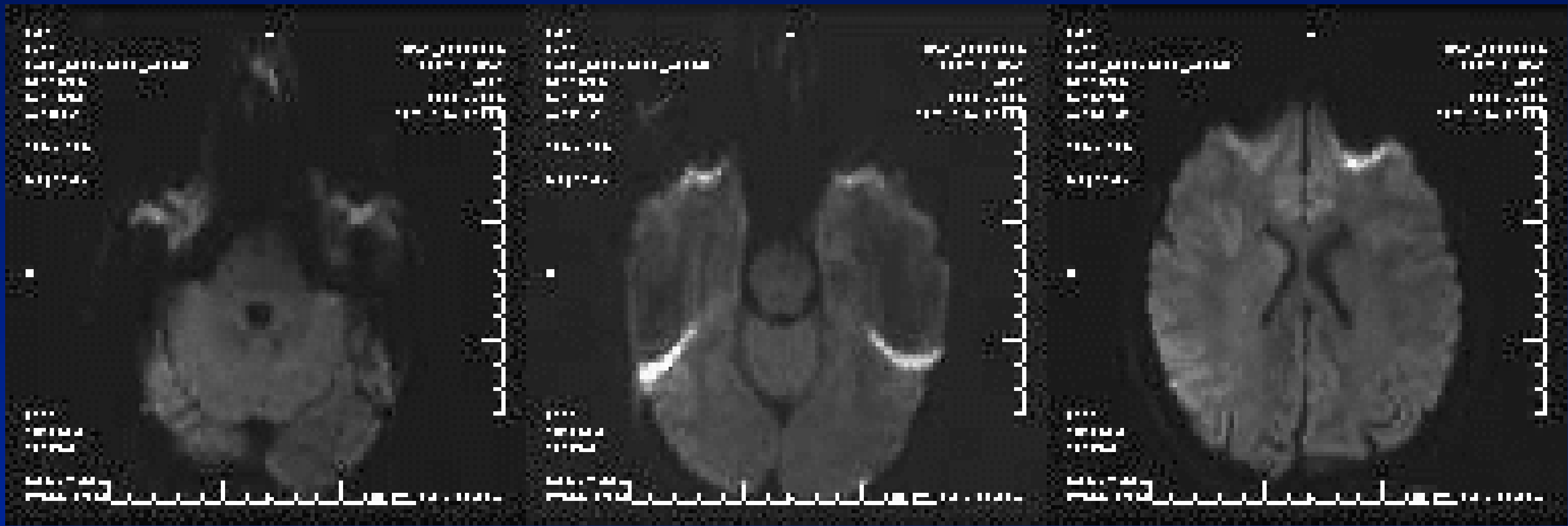
With fast gradients, add parallel imaging



(iPAT) GRAPPA for EPI susceptibility

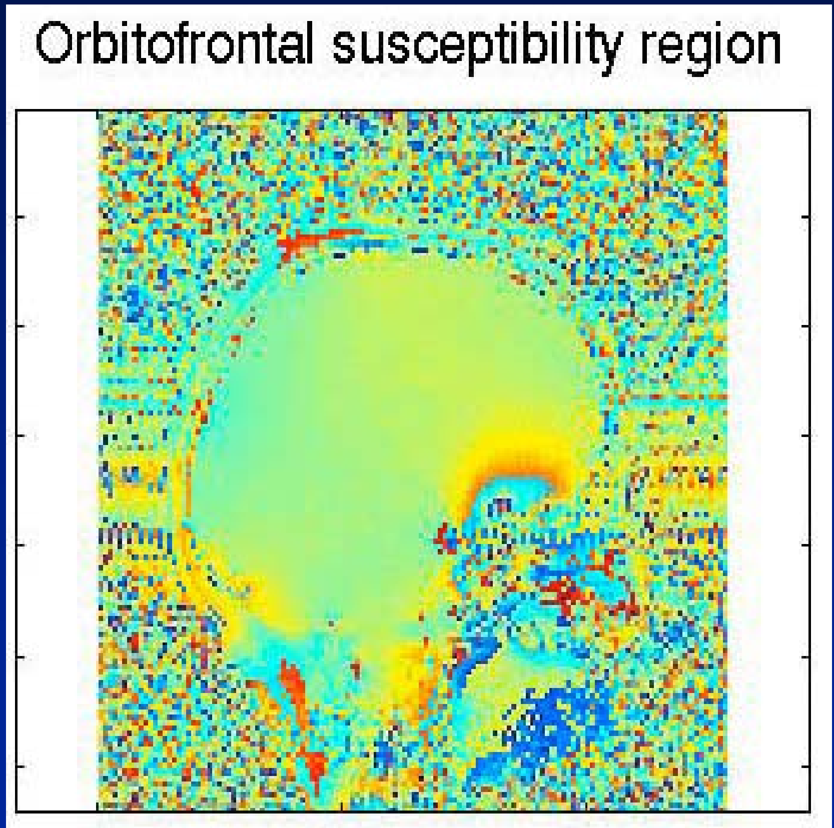
3T Trio, MRI Devices Inc. 8 channel array
b=1000 DWI images

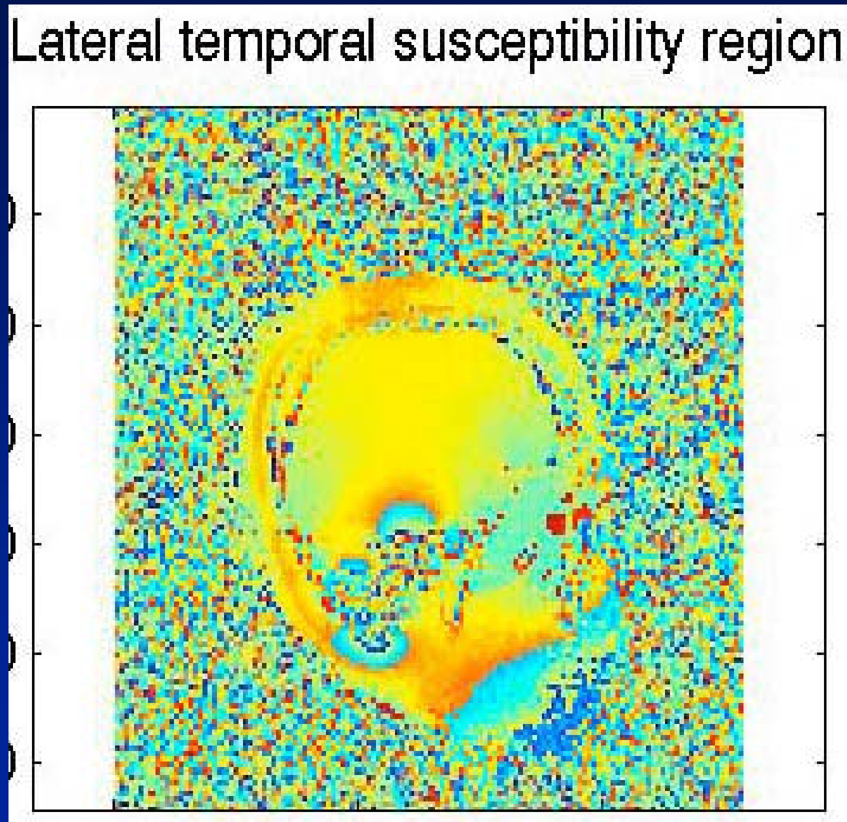


iPAT (GRAPPA) = 0, 2x, 3x

Fast gradients are the foundation, but EPI still suffers distortion

Enemy #1 of EPI: local susceptibility gradients





B_0 field maps in the head

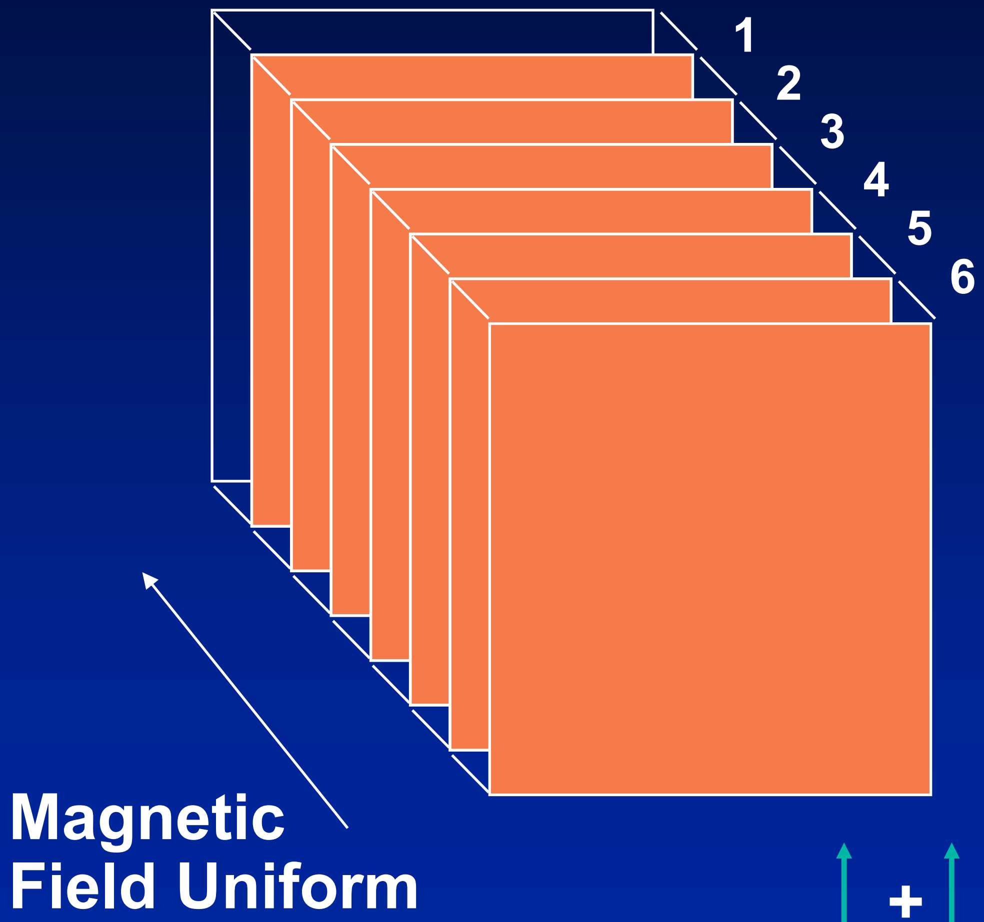
EPI: Local susceptibility gradients

Local susceptibility gradients have 2 effects:

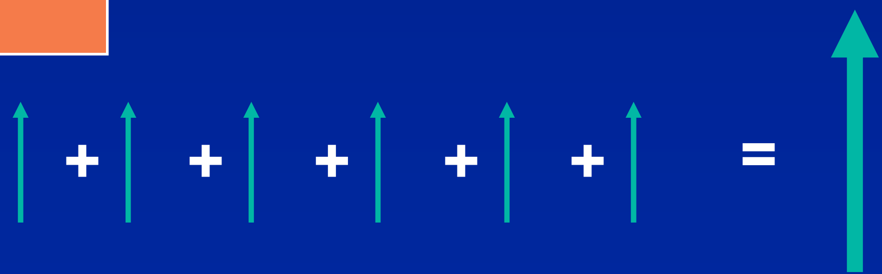
1) Local dephasing of the signal (signal loss) mainly from thru plane gradients

2) Local geometric distortions, mainly from local in-plane gradients.

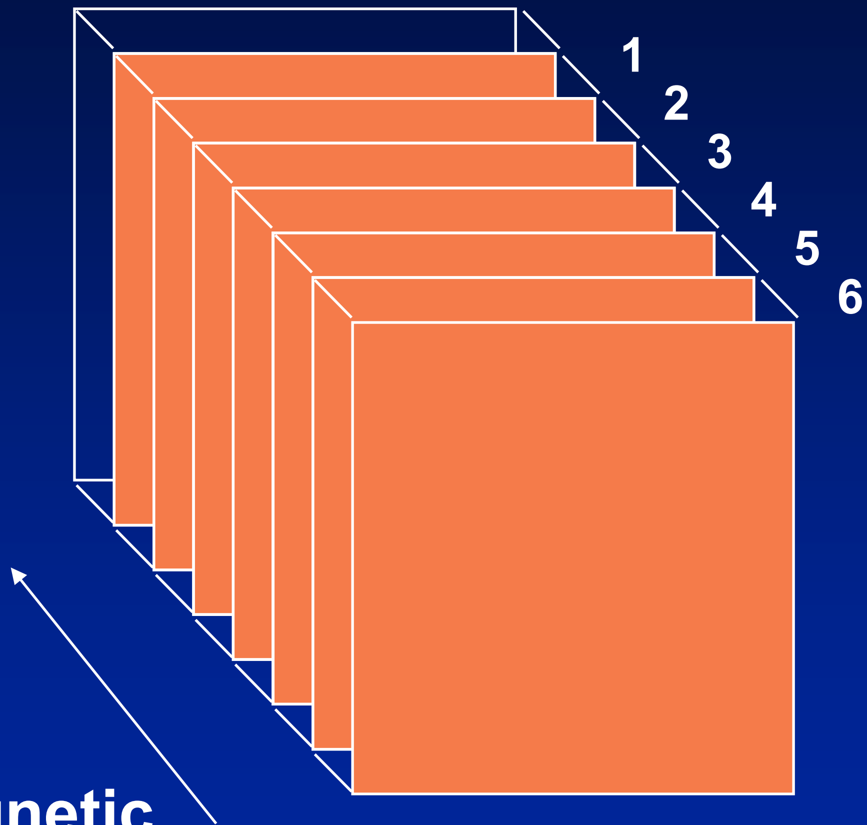
Susceptibility: thru plane dephasing



Signal from whole slice comes from adding together the MR vectors. When in phase, add constructively, SNR increases like slice thickness.

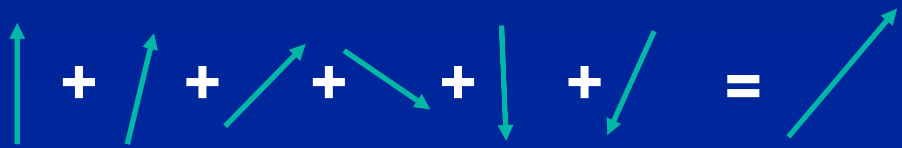


Susceptibility Artifact and Slice Thickness



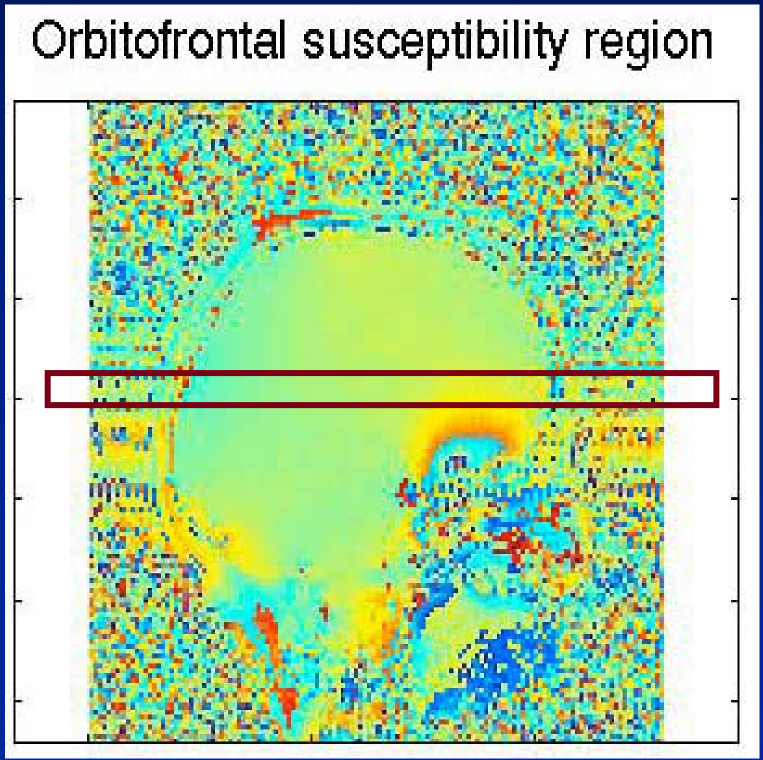
Magnetic Field NONUniform

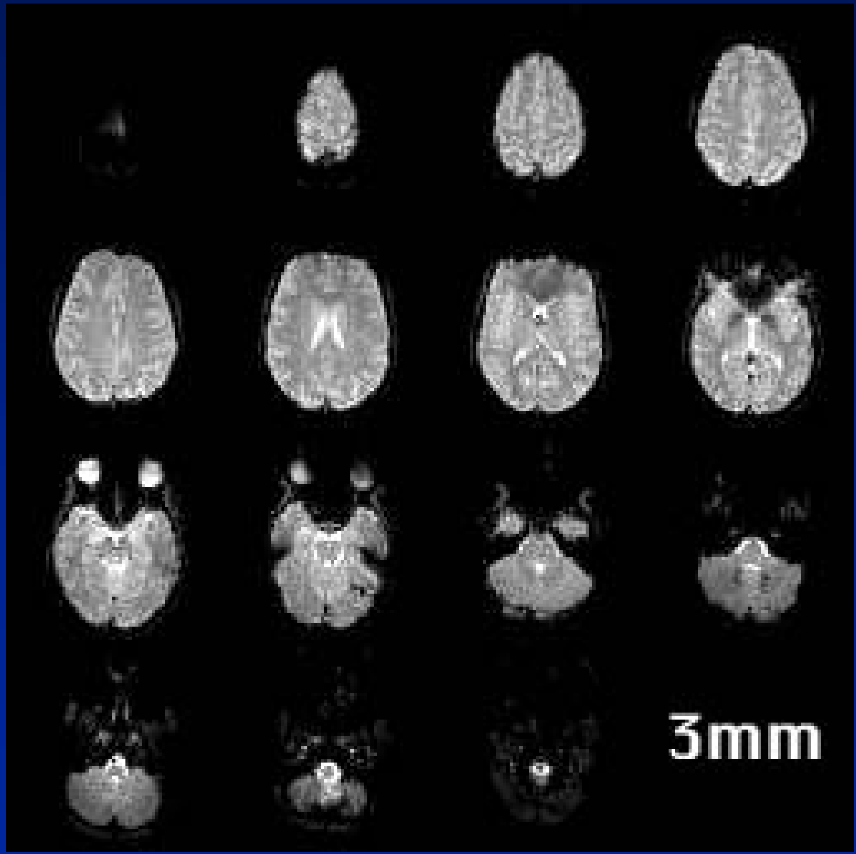
Signal from whole slice comes from adding together the MR vectors, which get out of phase when the magnetic field is not uniform



Local susceptibility gradients: thru-plane dephasing

Bad for thick slice above frontal sinus…





Local gradients: geometric distortion

Local gradient alters the helix of phase we have so carefully wound.

Phase error accumulates over entire kspace.
(conventional imaging phase is reset every line)

>> faster encoding is better.

Readout points are taken close together (~5us)

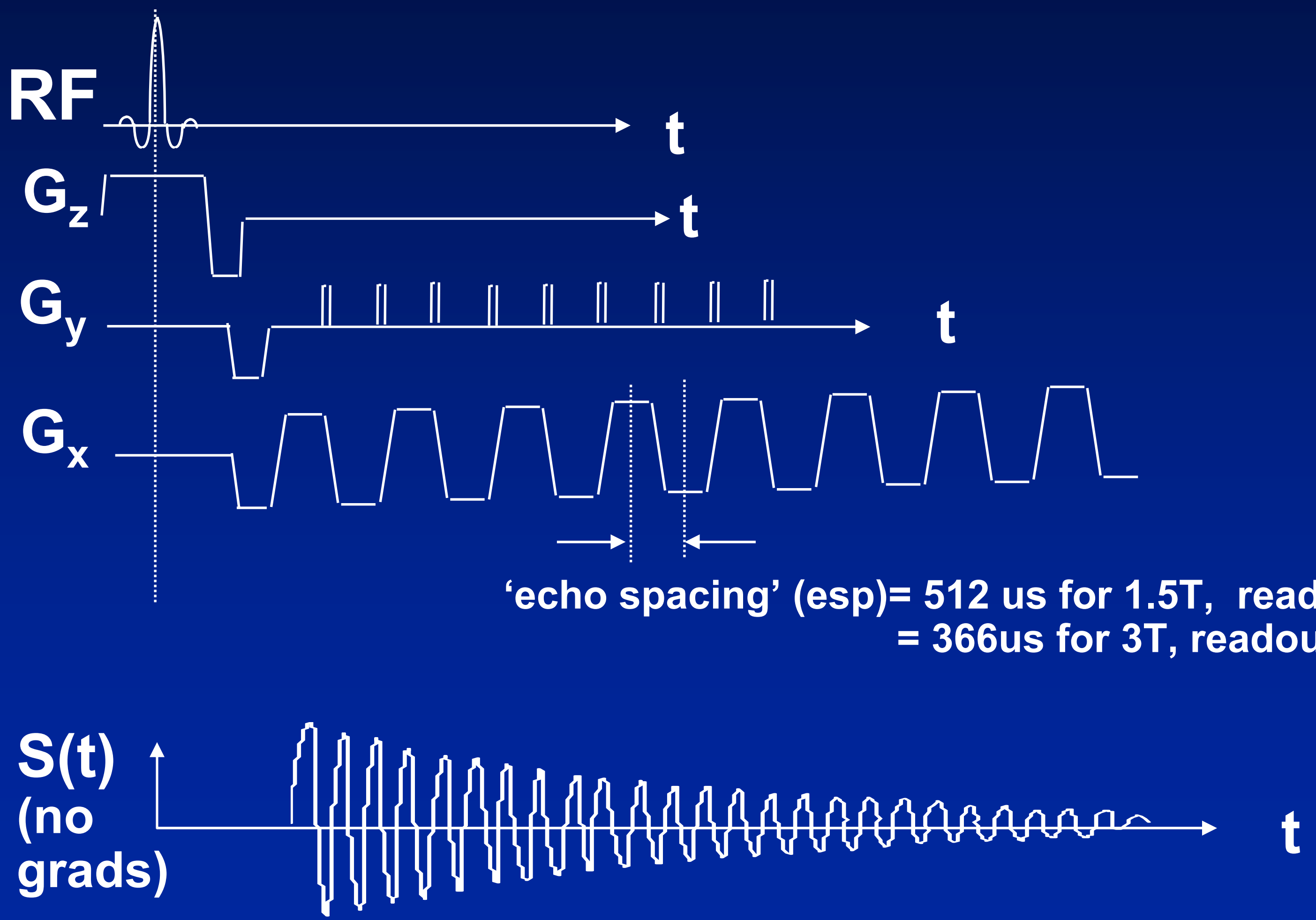
Phase encode points are taken farther apart (~500us)

>> distortion occurs in P.E. direction.

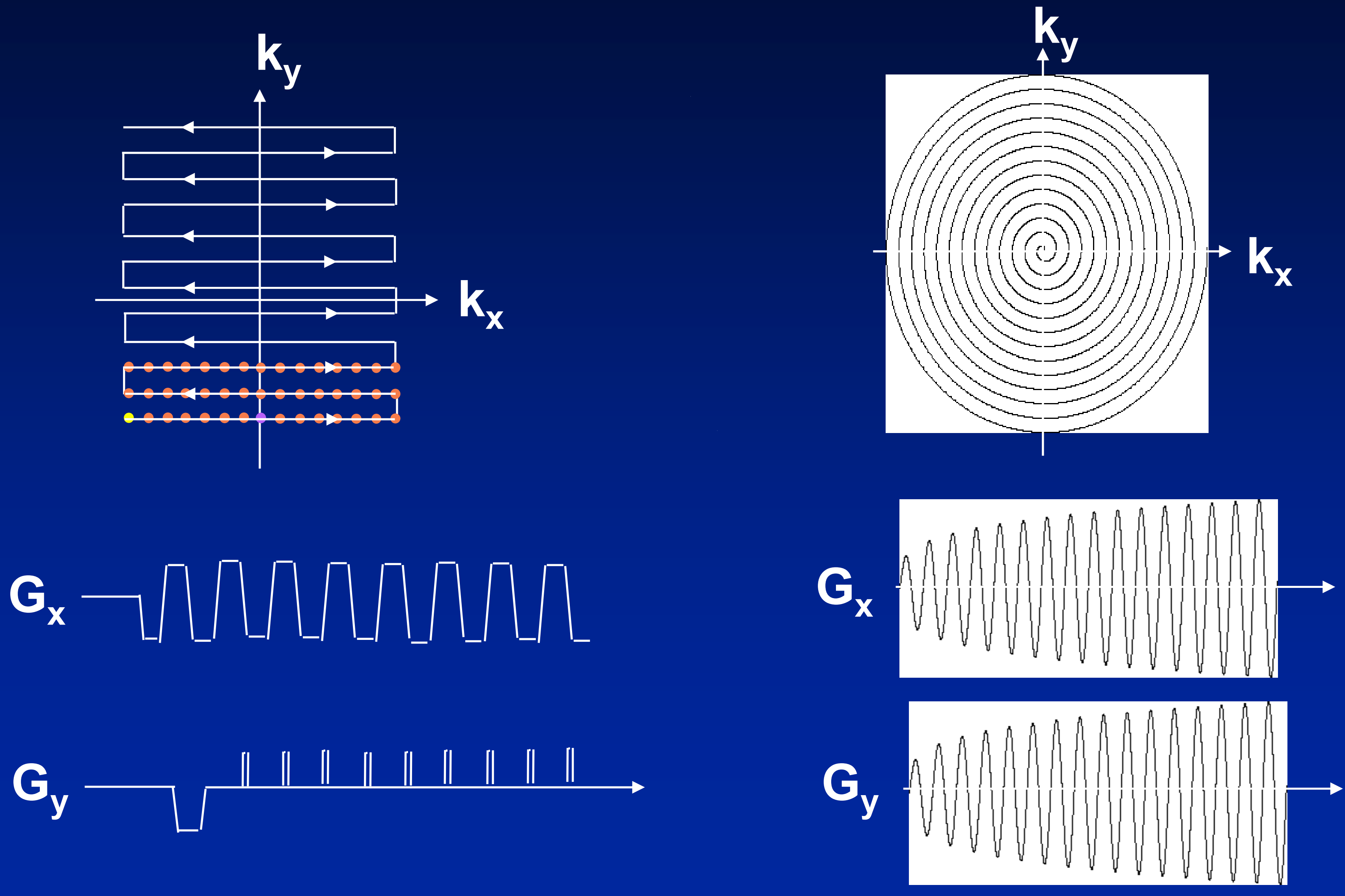
Characterization of grad. performance

- **length of readout train for given resolutio**

(requires fast slew and high grad amplitude)

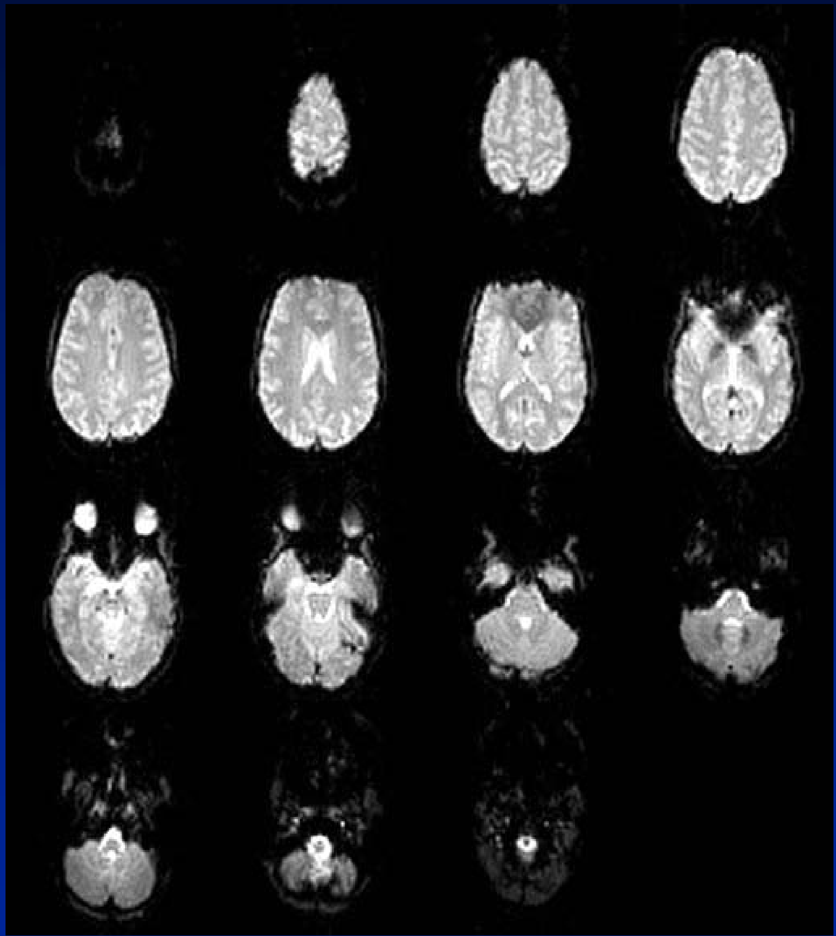


EPI and Spirals

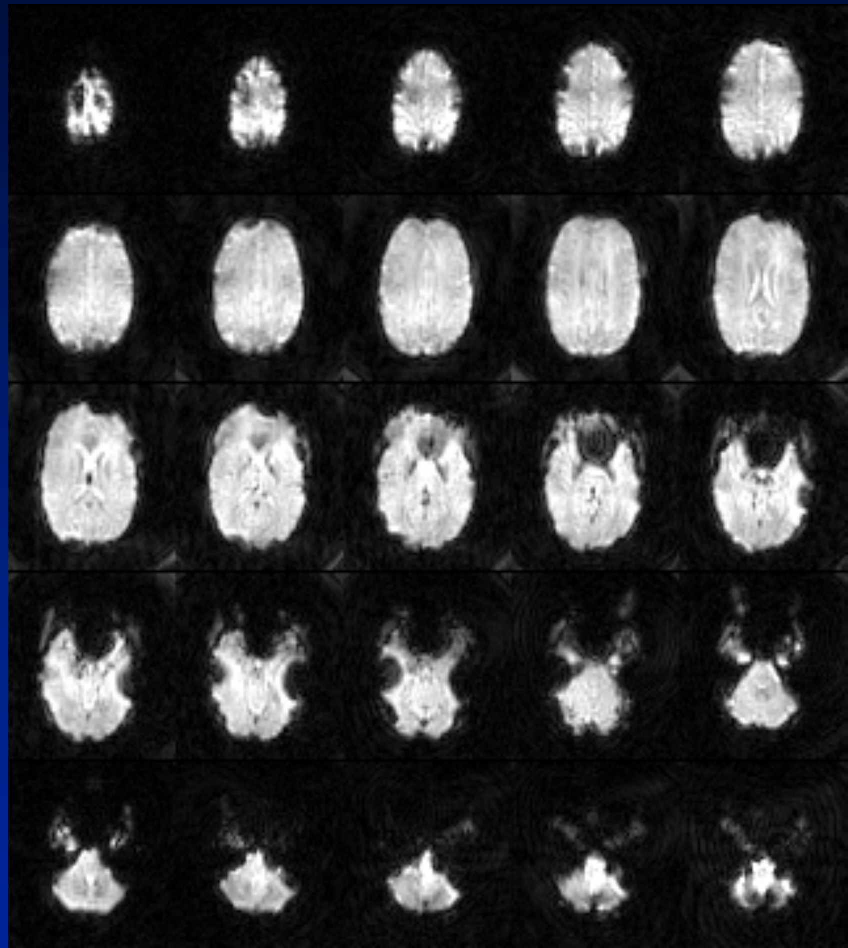


	EPI	Spirals
Eddy currents:	ghosts	blurring
Susceptibility:	distortion, dephasing	blurring dephasing
k = 0 is sampled:	1/2 through	1st
Corners of kspace:	yes	no
Gradient demands:	very high	pretty high

EPI and Spirals



EPI at 3T



Spirals at 3T
(from G. Glover)

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