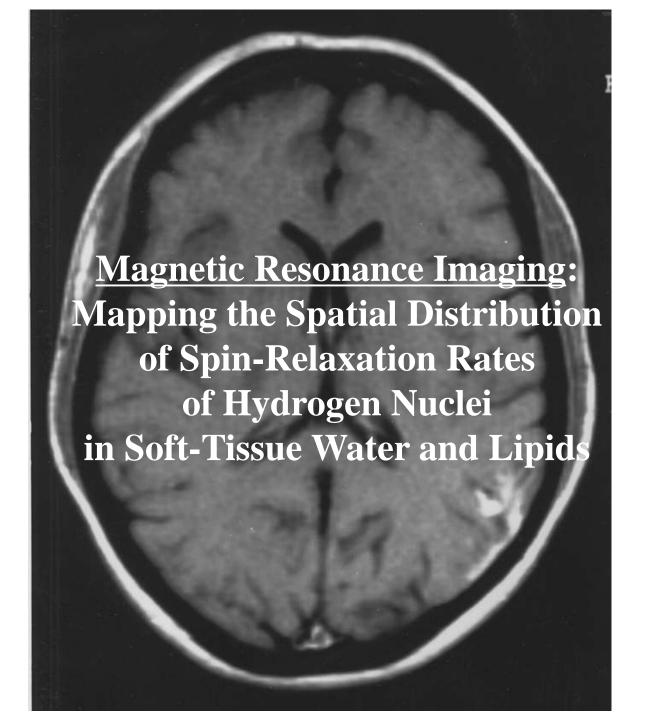
# A Brief Introduction to Magnetic Resonance Imaging

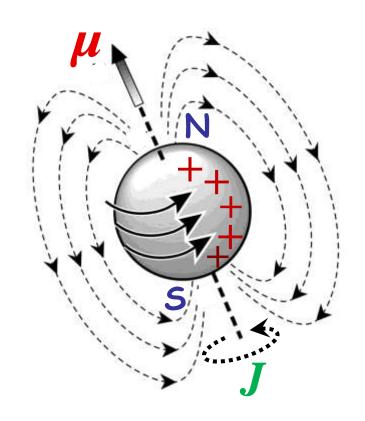


# Outline for Today

1)	Introduction MRI Case Study; <u>Caveat!!!</u> "Quasi-Quantum" NMR and MRI	Wolbarst
2)	Net magnetization, $m(x,t)$ , of the voxel at $x$ T1 Spin-Relaxation of $m(x,t)$ , T1 MRI of the 1D patient Sketch of the MRI Device	Lemen
3)	'Classical' Approach to NMR FID Image Reconstruction, <i>k</i> -Space	Yanasak
4)	Spin-Echo Reconstruction T2 Spin-Relaxation T1-w, T2-w, and PD-w S-MRI Spin-Echo / Spin-Warp in 2D Some Important Topics Not Addressed Here	Price



# Part 1: "Quasi Quantum" NMR and MRI

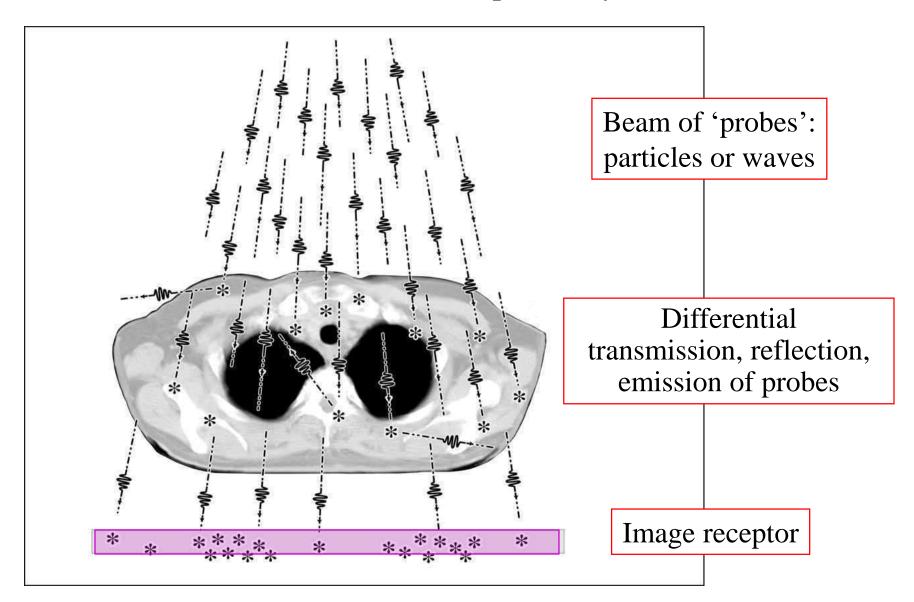


Introduction
MRI Case Study; *Caveat!!!*"Quasi-Quantum" NMR and MRI

# Introduction

#### **Medical Imaging:**

differential transmission, etc., of probes by different tissues

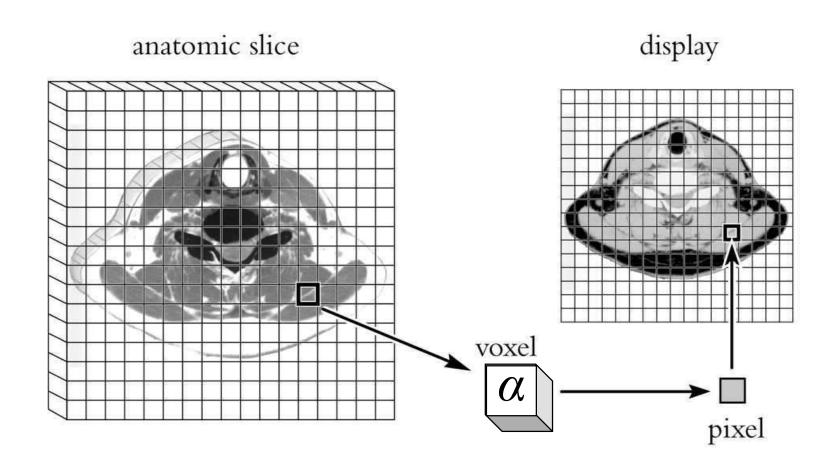


#### How Imaging Modalities Detect Tissue Contrast...

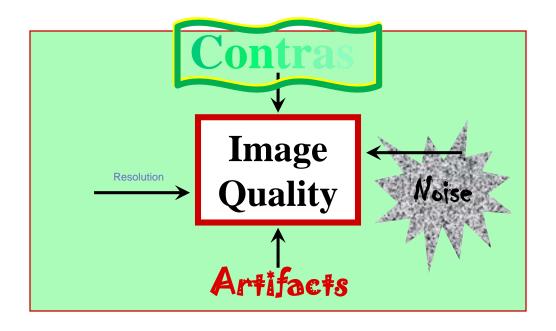
Modality	Probe / Signal	Detector	Source of Contrast: Differences in
Planar R/F X-ray CT	X-rays passing through body	II+CCD, AMFPI; GdO, etc., array	$\mu(\rho, Z, kVp), X$
Nuc Med, SPECT; PET	Gamma-rays from body; 511 keV	NaI single crystal; multiple NaI; LSO array	Radiopharmaceutical uptake, concentration ( <i>e.g.</i> , <sup>99m</sup> Tc), emission
US	MHz sound	Piezoelectric transducer	$ ho$ , $\kappa$ , $\mu_{\mathrm{US}}$
MRI	Magnet, RF	AM radio receiver	T1, T2, PD, [O], blood flow, water diffusion, chemical shift

#### Digital Display: Voxels to Pixels

tissue property parameterized by  $\alpha$  ( $\mu$ , [18FDG],  $\kappa$ ,T1,...)



#### Measures of Image Quality



#### Forms of Contrast Between Good and Bad Apples

produced and detected thru different bio/chemico/physical processes

**Color** 

Smell

Texture

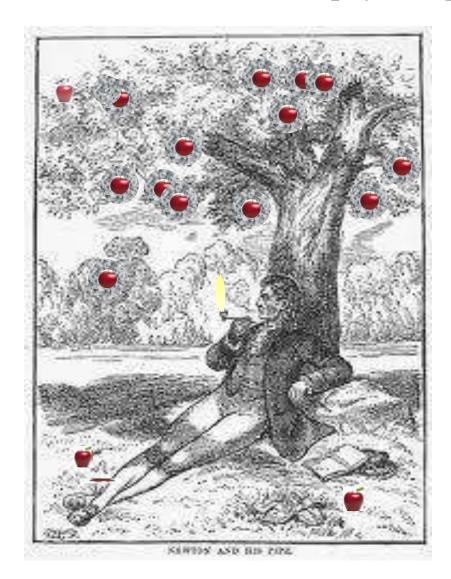
**H**oles

**Firmness** 

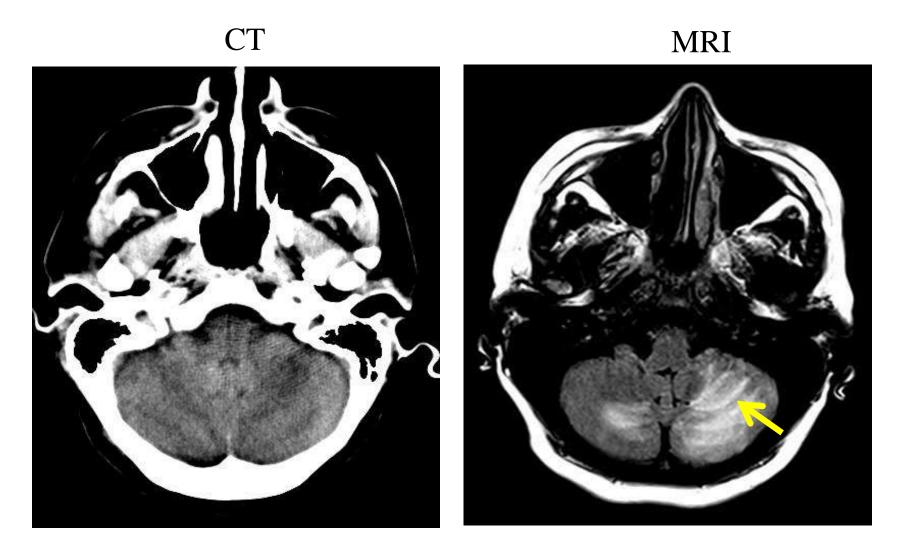
SHAPE

Taste

etc.



#### CT vs. MRI Soft Tissue Contrast



Posterior reversible encephalopathy syndrome (PRES), edematous changes

#### Magnetic Resonance Imaging

Multiple types of *contrast*, created through and reflecting different aspects of biophysics.

Most detect rotations, flow, *etc.*, of water/lipid molecules, yielding unique maps of tissue anatomy, physiology: *vessels, nerve trunks, cartilage, tumors,...* 

No ionizing radiation

Risks from intense magnetic fields, RF power

Expensive

Technology complex, challenging to learn

# MRI Case Study; and Caveat!!!

with T1, FLAIR, MRS, DTI, f MRI, and MR-guided biopsy studies of a glioma

#### Case Study

57 year old ♀ medical physicist has had daily headaches for several months. Responds fully to ½ Advil.

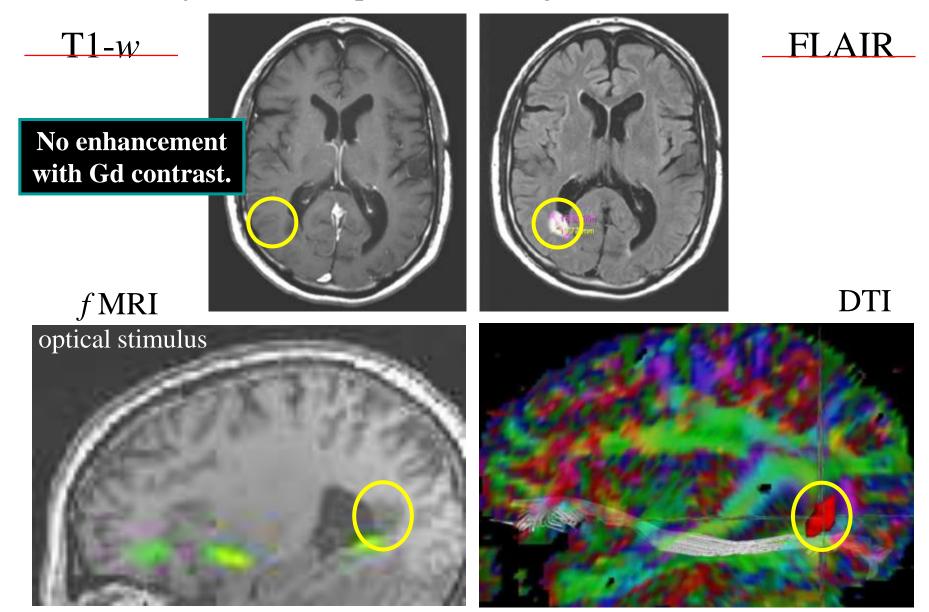
Physical examination at local clinic unremarkable. Patient appears to be in good general health, apart from mild hypertension, controlled by medication.

Good diet, exercises moderately. Patient reports no major stresses, anxieties.

CT indicates a lesion in the *right posterior temporo-occipital* region, adjacent to *occipital horn of right lateral ventricle*. MRI for better contrast.

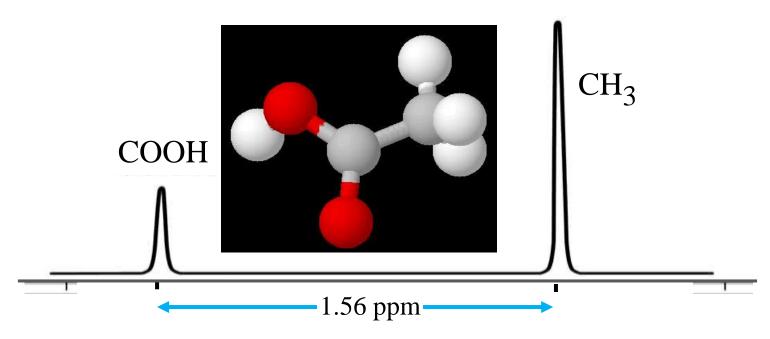
Principal concern: Vision for reading.

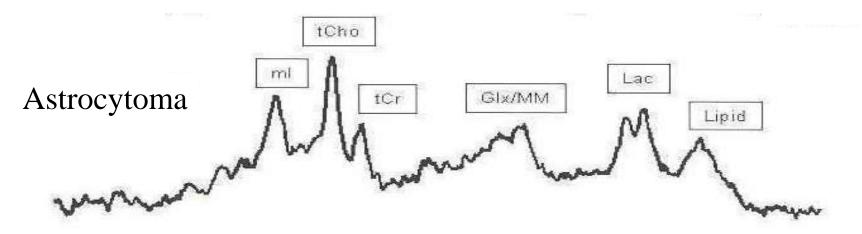
# Right Posterior Temporo-Occipital Region, adjacent to occipital horn of right lateral ventricle



#### Chemical Shift and Non-invasive MRS

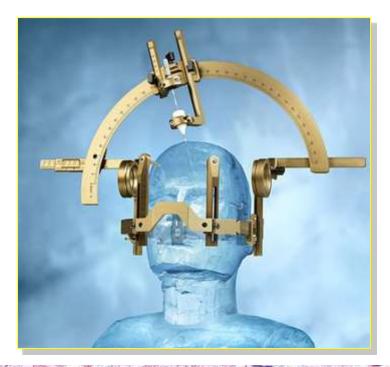
$$f_{\text{Larmor}} \text{ (MHz)} = 42.58 (1 + \sigma) B_0 \text{ (T)}$$





#### MRI-Guided Needle Biopsy







# Caveat!!!

Extreme Danger

## Strong Gradient Fields Outside Bore



#### How Not to Clean the Device



#### MRI Safety

fringes of principal magnetic field rapidly switching (gradient) fields high RF power

FDA: ~ 40 MRI-related accidents/year 70% RF burns, 10% metallic "missiles"

RF Specific Absorption Rate (SAR): dE/dm (W/kg)

no magnetic *anything* in or entering MRI room restricted *access*, safe zones, prominent warnings accompany *all* patients, visitors, non-MRI staff accurate medical *history*; double-check for metal training for *any* staff (*e.g.*, cleaners)

possible *gadolinium* risk: nephrogenic systemic fibrosis (NSF) *pregnant* patients generally should not have Gd-contrast

#### MRI Safety – Prevent Entry!

within / with patient, others

aneurysm clip, shrapnel cochlear implant, prostheses artificial heart valve stent, permanent denture defibrillator, pacemaker, electrodes, nerve stimulator medical infusion pump drug-delivery patch, tattoo

in I into imaging suite

hemostat, scalpel, syringe  $O_2$  bottle, IV pole scissors, pen, phone, laptop tool, tool chest wheelchair, gurney ax, fire extinguisher gun, handcuffs cleaning bucket, mop

## "Quasi-Quantum" NMR and MRI

Swept-frequency NMR in a tissue voxel Proton-Density (PD) MRI in the 1D patient Magnetization, m(x,t), in the voxel at x

#### Two Approaches to Proton NMR/MRI (incompatible!)

quantum spin-state function for hydrogen nucleus

## start with this

Simple QM  $|\uparrow\rangle$ ,  $|\downarrow\rangle$ 

transitions between spin-up-, spin-down states

 $f_{\text{Larmor}}$ ,  $m_0$ , T1

oversimplified



Classical Bloch Eqs. expectation values in voxel

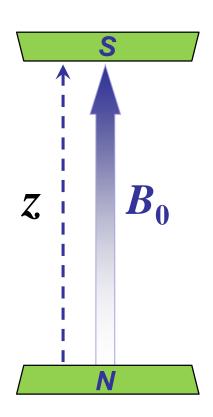
precession, nutation of net voxel magnetization, m(t)

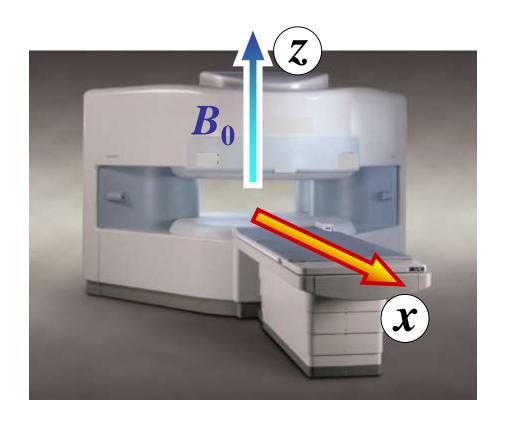
 $f_{\text{Larmor}}$ , T2, 2D, k-space

exact; from full QM

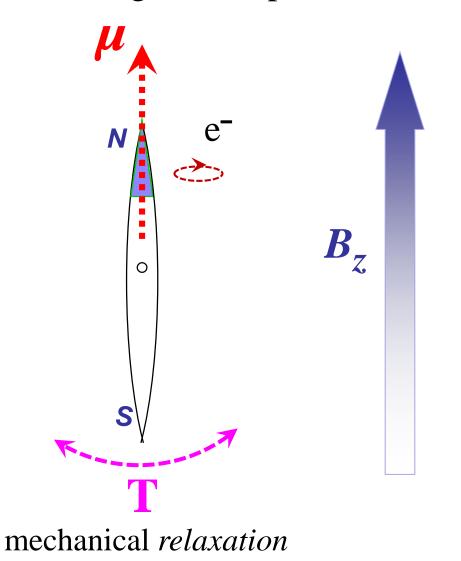
#### 'Open' MRI Magnet

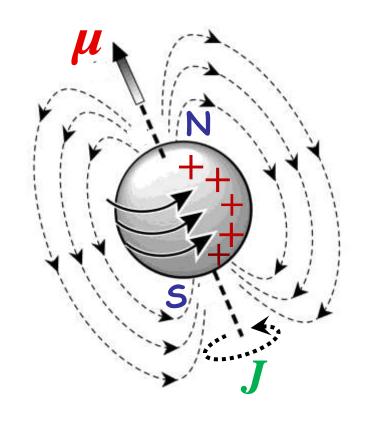
#### principal magnetic field $\boldsymbol{B}_0$ defines $\boldsymbol{z}$ -axis





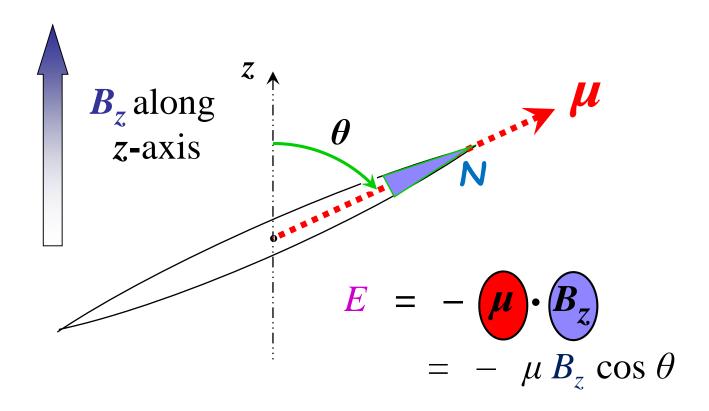
# Moving Charge Produces Magnetic Field (e.g., mag. dipole); Magnetic Dipole Tends to Align in External Field





simplified QM picture *proton* 

## Energy to Flip Over Needle with Magnetic Moment $\mu$ in $B_z$

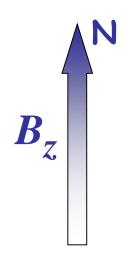


$$\Delta E_{180^{\circ}} = \pm 2 \,\mu \,B_{z}$$

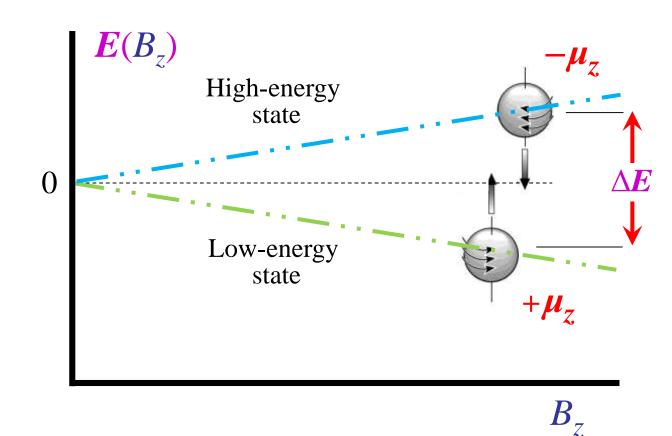
## Swept-frequency NMR in a tissue voxel

# Nuclear Zeeman Splitting: $\Delta E = \pm 2 \mu_z B_z$ points only along or against z

#### South



North



#### Nuclear Magnetic Resonance

$$\begin{cases} \Delta E_{180^{\circ}} = 2 \mu_z B_z \\ \text{photon } E = hf \end{cases}$$

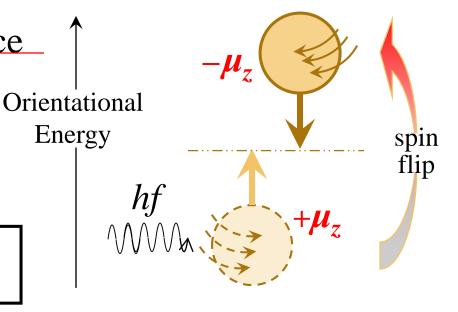
$$f_{\text{Larmor}}(B_z) = (2/h)\mu_z B_z$$

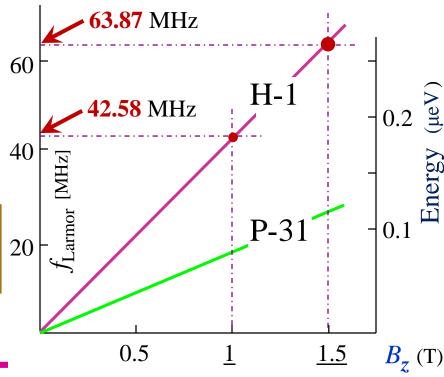
$$2\pi f_{\text{Larmor}} \equiv \omega = \gamma B_z$$

for water protons,

$$f_{\text{Larmor, H}_2\text{O}} = 42.58 \left[ \frac{\text{MHz}}{\text{T}} \right] \times B_z \left[ \text{T} \right] 20$$

$$B_z(x) = f_{Larmor}(x) / 42.58$$





#### Proton Spin Transitions: µeV



therapy x-rays, electrons: MeV

radioactivity: keV – MeV

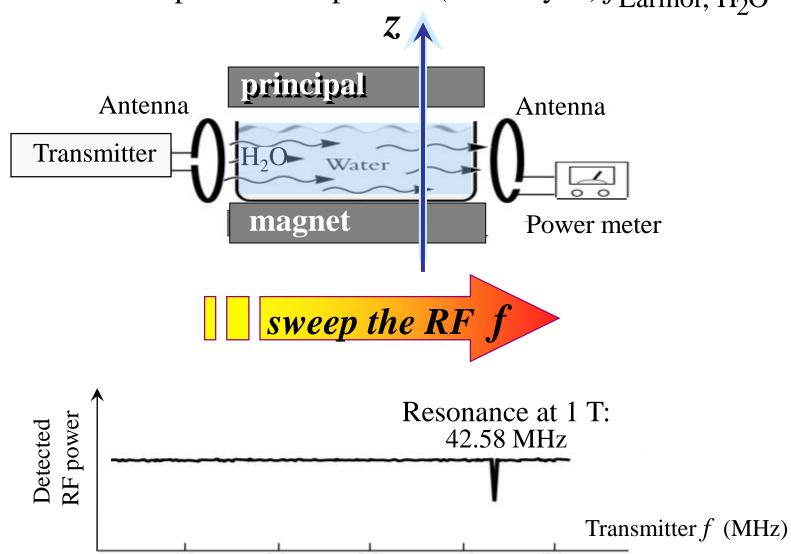
diagnostic x-rays: 30 - 120 keV

visible light, chemical reactions: eV

MRI: μeV (~100 MHz)

#### NMR Gedanken-Experiment on Water

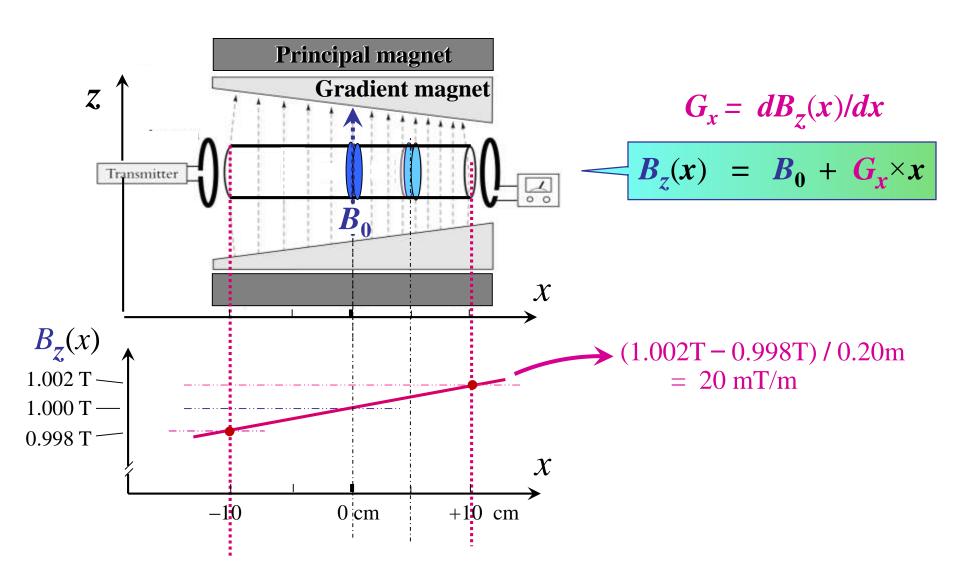
monochromatic RF power absorption at (and only at)  $f_{\text{Larmor, H}_2\text{O}}$ 

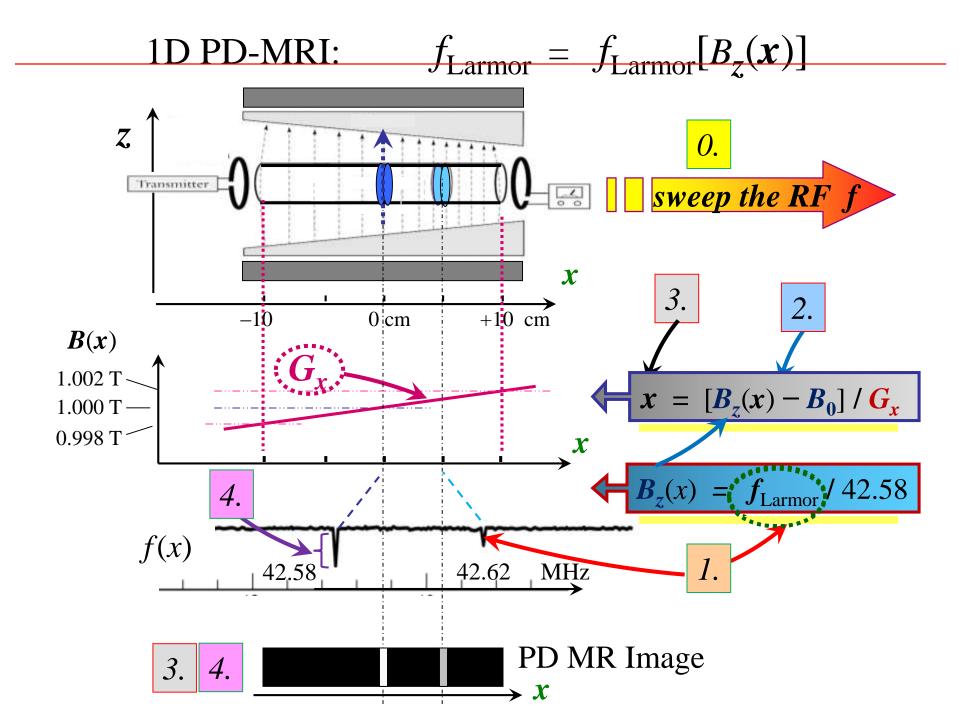


## Proton-Density (PD) MRI in the 1D patient

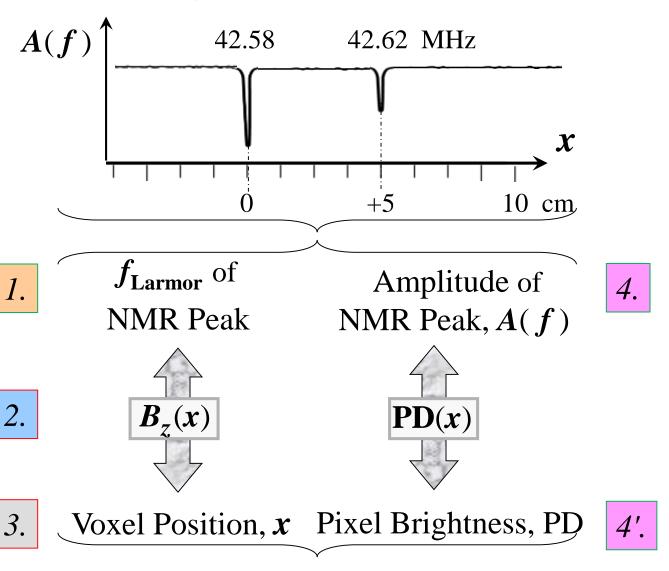
## 1D Phantom in Principal $B_0$ and Gradient $G_x$

$$x = [B_{z}(x) - B_{0}] / G_{x}$$





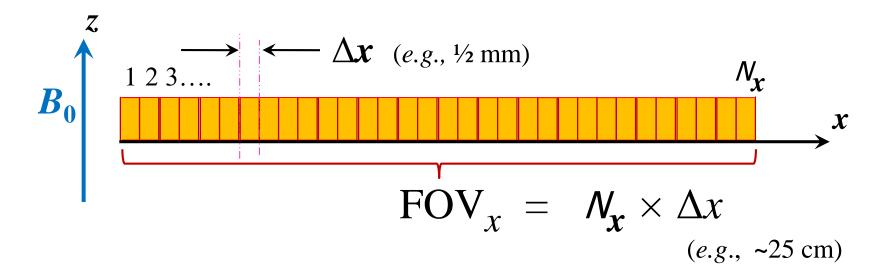
#### Summary: PD MRI on 1D Patient





### Matrix Size, Resolution, and Field of View

$$N_{\chi}$$
 = voxels (matrix dimension) along *x*-axis (e.g., 256)

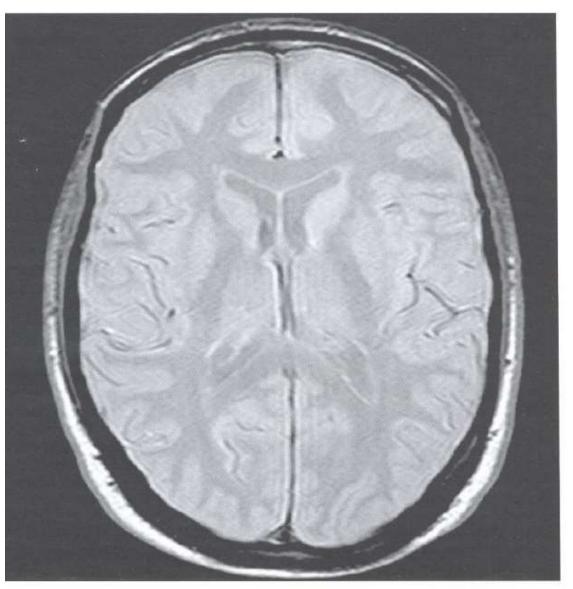


In-plane resolution:  $2\Delta x \sim 1 \text{ lp/mm} \sim 1 \text{ mm}$ (but may see object  $\leq \Delta x$ )

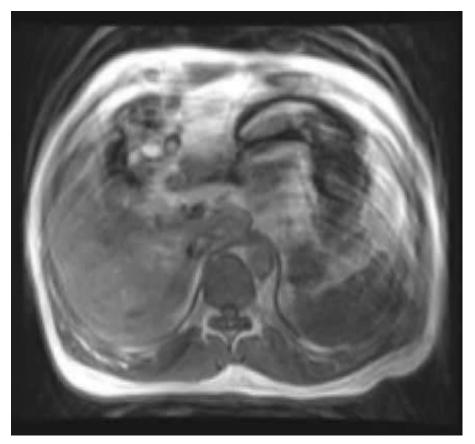
Imaging plane: ~ 1 mm thick.

### Proton-Density MRI

### contrast from differences in PD



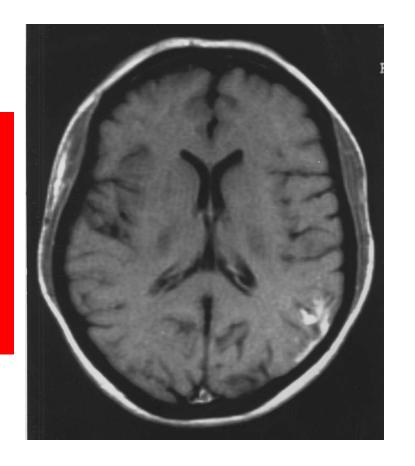
### Two MRI Motion Artifacts



respiration

aortic pulsation

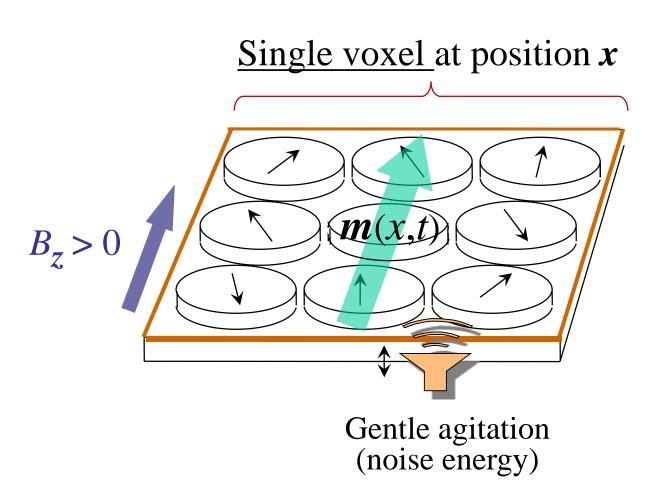
# Part 2: Magnetization & Relaxation



Net magnetization, m(x,t), of the voxel at x T1 Spin-Relaxation of m(x,t), T1 MRI of the 1D Patient Sketch of the MRI Device

Net magnetization, m(x,t), of the voxel at x

# Net Magnetization, m(x,t), for the <u>Single Voxel</u> at Position x: magnetic field *from* the ensemble of protons or needles *themselves*

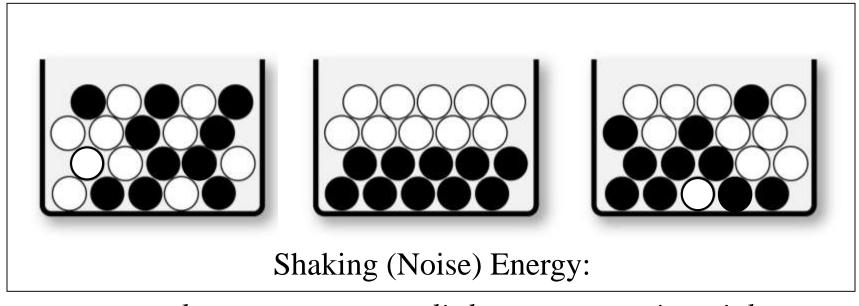


### Voxel's MRI Signal Proportional to Its Magnetization, m(x,t)

- i) What is the magnitude of voxel magnetization at dynamic thermal equilibrium,  $m_0$ ?
- ii) How long does it take to get there (T1)?
- iii) What is the mechanism?

# Filling Four Energy Levels of Marbles vs. Noise Level battle between energy and entropy

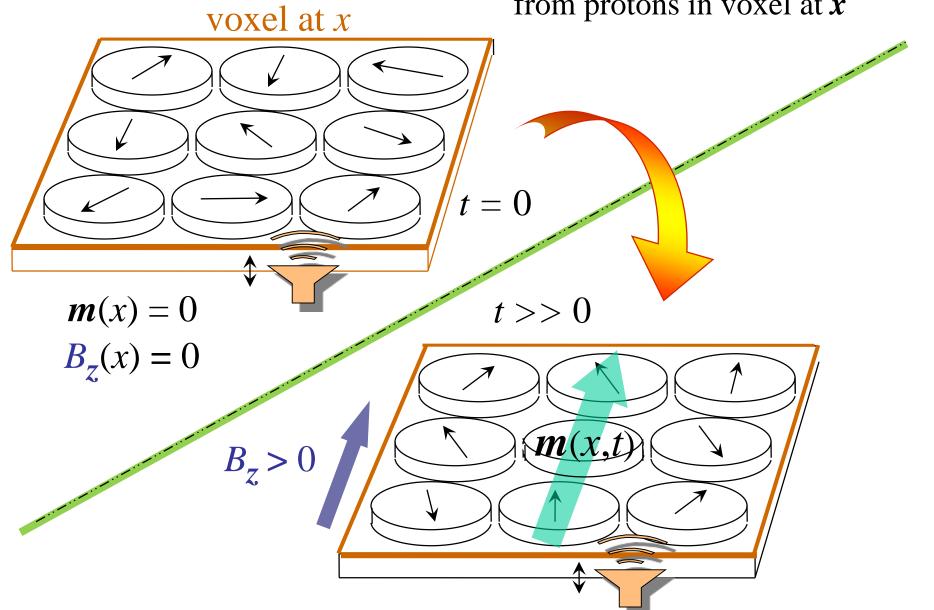
(all slippery; black balls denser)



too much too little just right

### Switching $B_z(x)$ On at t = 0 Induces Magnetization m(x,t)

from protons in voxel at x

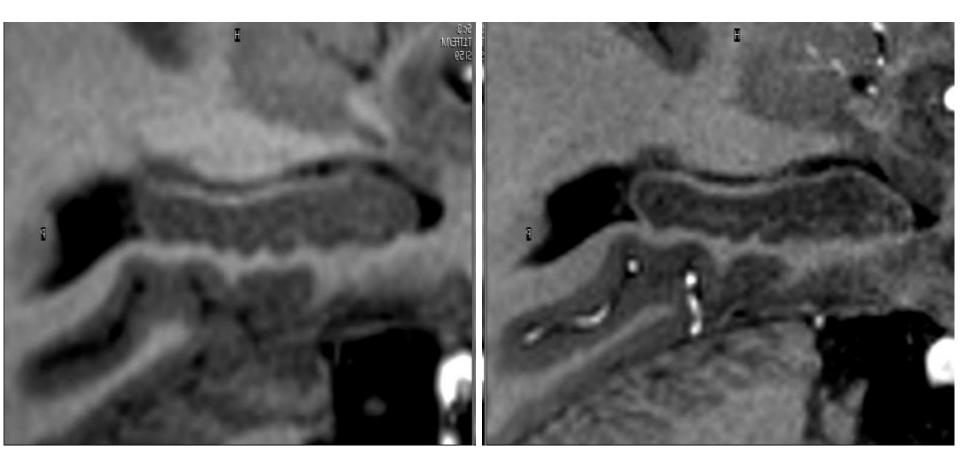


Magnetization in Voxel at x:  $m(x,t) = [N_{-}(x,t) - N_{+}(x,t)] \times \mu$  collective magnetic field produced by all the protons in it

_ B <sub>z</sub>	$N \equiv (N_+ + N_+)$ protons/voxel	(Boltzmann) <b>m</b>
0  tesla	STO STO STO	0
$\begin{array}{c c} 0.01 \text{ T} \\ or \\ t = 0 + * \end{array}$		~0
»1.5 T		$N\mu_z$
$\begin{vmatrix} 1.5 \text{ T} \\ t >> 0 \end{vmatrix}$		5×10 <sup>-6</sup> Nµ <sub>z</sub>

<sup>\*</sup> after abruptly turning on  $B_z$ , or after a 90° pulse.

 $|m_0(x)|$  and Signal from it Increase with  $B_0$  p.s., trade-off: SNR, resolution, acquisition-time



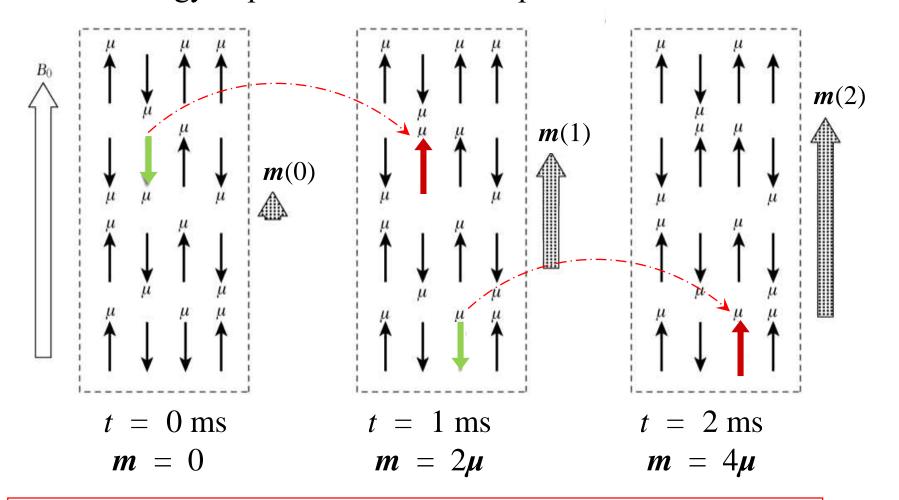
1.5 T 7 T (+Gd)

### T1 Spin-Relaxation of m(x,t)

### Voxel's MRI Signal Proportional to Its Magnetization, m(x,t)

- i) What is the magnitude of voxel magnetization at thermal equilibrium,  $m_0$ ?
- ii) How long does it take to get there (T1)?
- iii) What is the mechanism?

# Disturbed System Moving toward Up-Down Equilibrium energy imparted to individual spins from 'outside'



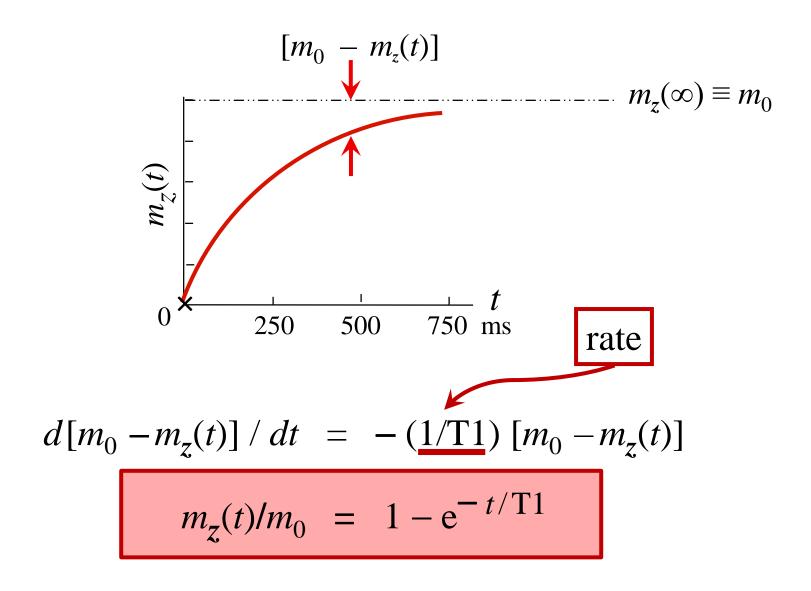
spin 'tickled' down by  $f_{\text{Larmor}}$  component of magnetic noise; emits RF photon, phonon

### Decay of Tc-99m Sample to Final Value, $n(\infty) = 0$

n(0) radionuclei at initially n(t) remaining after time t.  $n(\infty)$  at 'equilibrium'  $n(\infty) = 0$  $dn/dt(t) = -\lambda n(t)$  $n(t) = n(0) e^{-\lambda t}$ 1/2  $dn/dt \sim [n(t) - n(\infty)]$ 

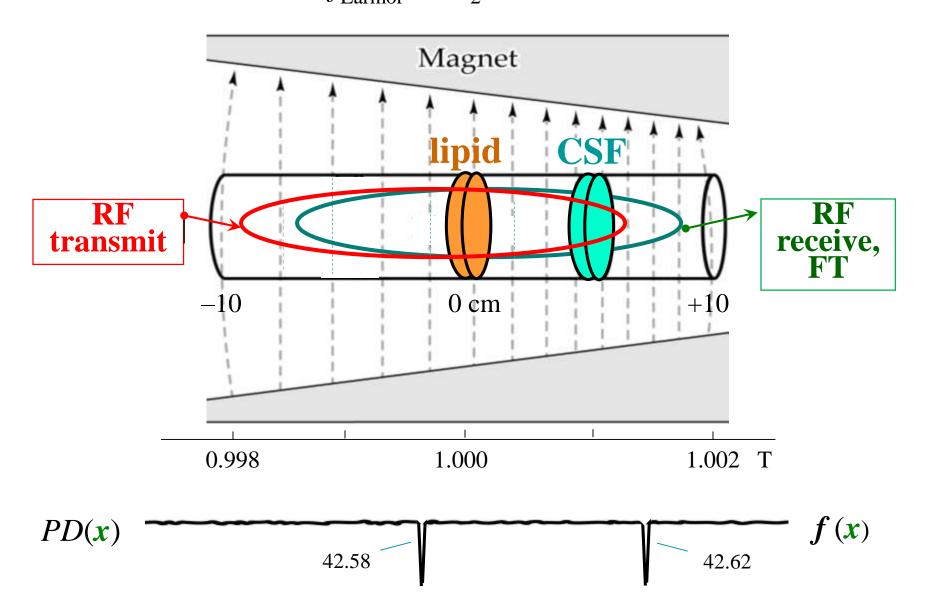
scint. decay 
$$(t)$$
 pop. growth  $(t)$  photon atten.  $(x)$  tracer conc.  $(t)$  cell killing  $(D)$  ultrasound atten.  $(x)$ 

### Return of Voxel's $m_z(t)$ to Final Value, $m_z(\infty) \equiv m_0$

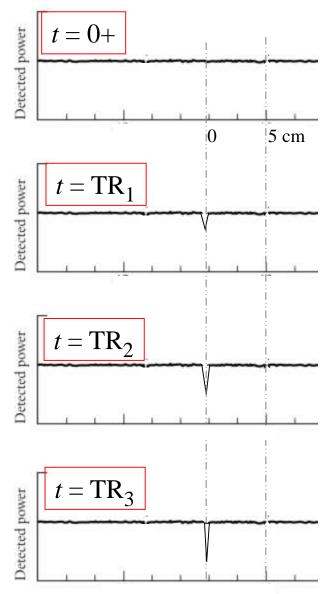


### T1 MRI of the 1D Patient

T1 for 2 Voxels at 1 tesla: Lipid vs. CSF both at  $f_{\text{Larmor}}$  of H<sub>2</sub>O for local field



# Recovery of $m_z(t)$ over Time: $m_z(TR_i)$ Measurements longitudinal component of net voxel magnetization, $m_z(t)$

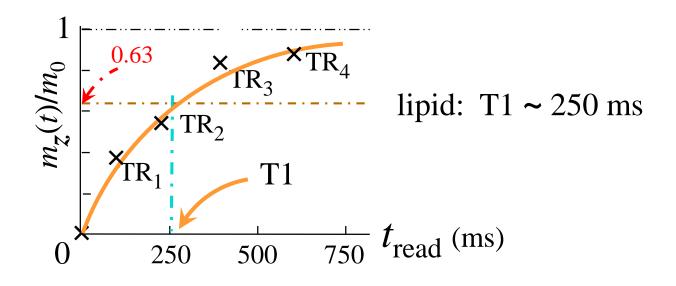


- 1. First, do *lipid* at x = 0.
- 2. Strong burst of RF over range  $42.58 \text{ MHz} \pm 250 \text{ Hz}$  causes



- 3. After  $TR_1$  delay, sweep through  $f_{Larmor}$ , record amplitude.
- 4. Repeat for several TR<sub>i</sub> values.
- 5. Plot.

$$m_z(TR)/m_0 = (1 - e^{-TR/T1})$$
  
curve-fit for set of  $\{TR_i\}$ 

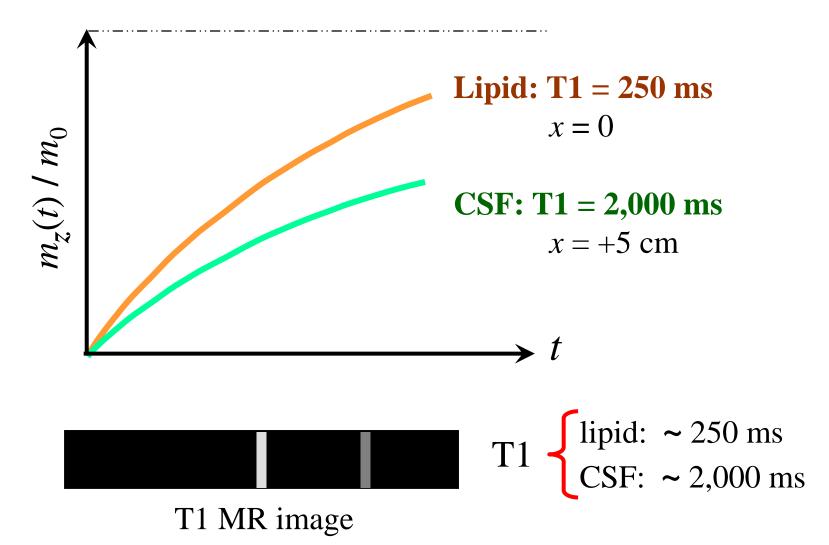


T1 MR image

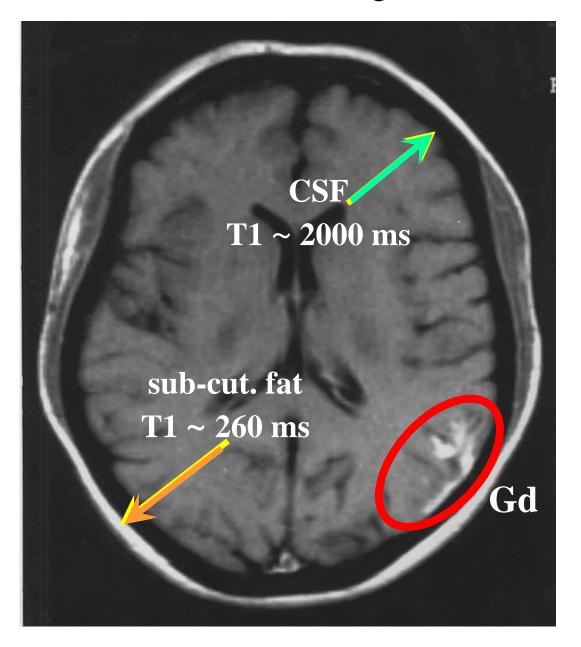
by convention for T1-imaging: Spritely Brightly

$$m_z(x,t)/m_0(x) = (1 - e^{-t/T1})$$

repeat for CSF voxel at x = 5 cm,  $f_{Larmor} = 42.62$  MHz...



### T1-w MR Image



### Approximate Relaxation Times of Various Tissues

Tissue	PD p+/mm <sup>3</sup> , rel.	T1, 1T (ms)	T1, 1.5T (ms)	T1, 3T (ms)	T2 (ms)
pure H <sub>2</sub> 0	1	4000		4000	4000
brain					
CSF	0.95	2000	2000	2000	200
white matter	0.6	700	800	850	90
gray matter	0.7	800	900	1300	100
edema			1100		110
glioma		930	1000		110
liver			500		40
hepatoma			1100		85
muscle	0.9	700	900	1800	45
adipose	0.95	240	260		60

### Voxel's MRI Signal Proportional to Its Magnetization, m(x,t)

- i) What is the magnitude of voxel magnetization at thermal equilibrium,  $m_0$ ?
- ii) How long does it take to get there (T1)?
- iii) What is the mechanism?

In MRI, the <u>only</u> thing a proton is <u>ever</u> aware of, or reacts to, is the <u>local magnetic field</u>,  $B_{local}(t)$ .

!!!

But the source of  $f_{Larmor}(B_{local}(t))$  can be either <u>external</u>  $(B_{RF})$  or <u>internal</u> (e.g., moving partner)

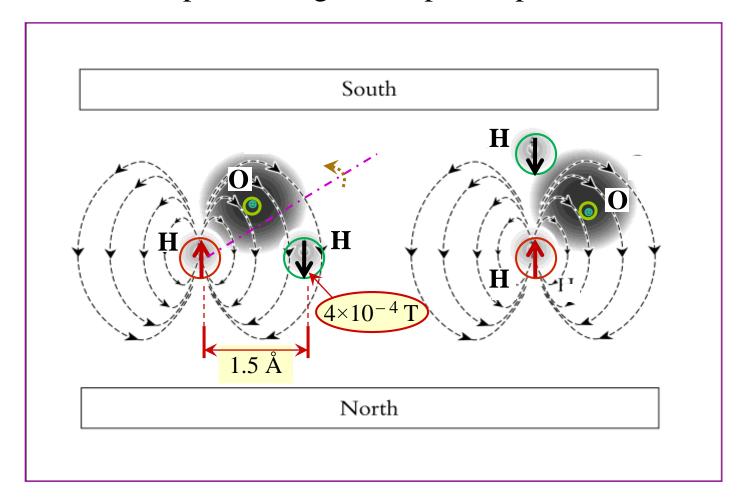
### Nuclear Magnetic Resonance vs. T1 Relaxation!

$$f_{\text{Larmor}} = 42.58 B_{z}(x)$$

<u>NMR</u>: Magnetic field of Larmor-frequency *photon* can couple to *magnetic dipole moment* of proton in lower-energy spin state, transferring energy  $hf_{Larmor}$  to it, flipping it "up."

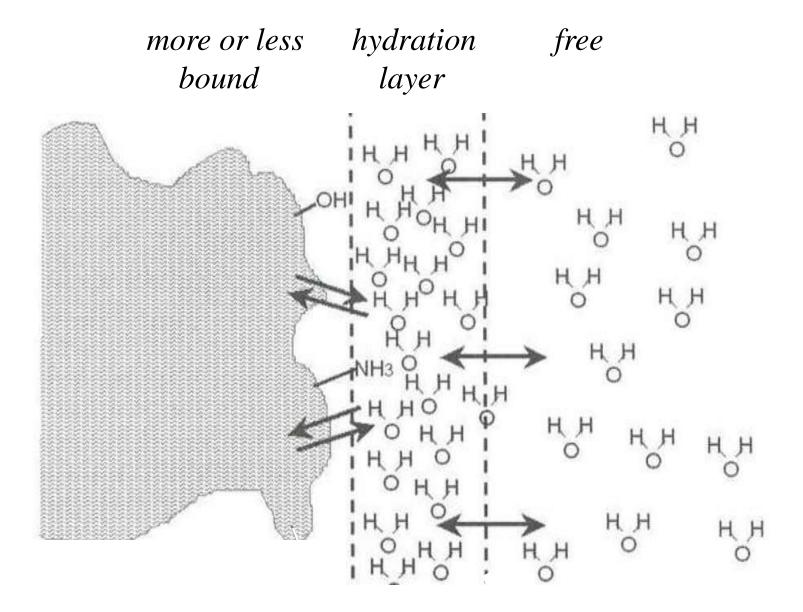
<u>T1 Relaxation</u>: Any magnetic field varying at  $f_{Larmor}$  can 'tickle' a proton down from the higher-energy spin state, with the proton giving off a photon (RF) or phonon (heat) of energy  $hf_{Larmor}$ .

## Spontaneous $f_{Larmor}$ - $B_{RF}(t)$ Causes T1-Transitions variations in proton magnetic dipole-dipole interactions

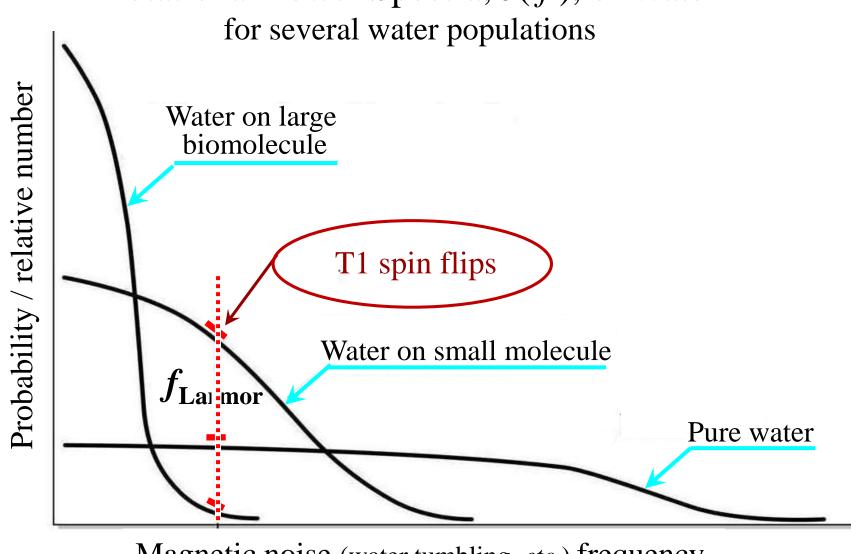


each water proton produces magnetic field fluctuations of various frequencies, including local  $f_{Larmor}$ , at its partner proton

### Restrictions on Rotations of Water Molecules



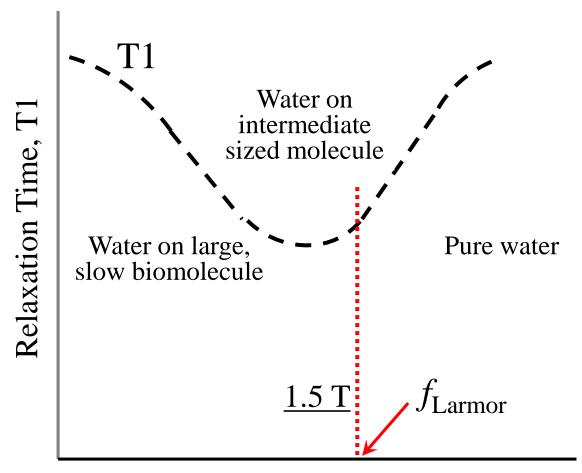
### Rotational Power Spectra, J(f), of Water



Magnetic noise (water tumbling, *etc.*) frequency

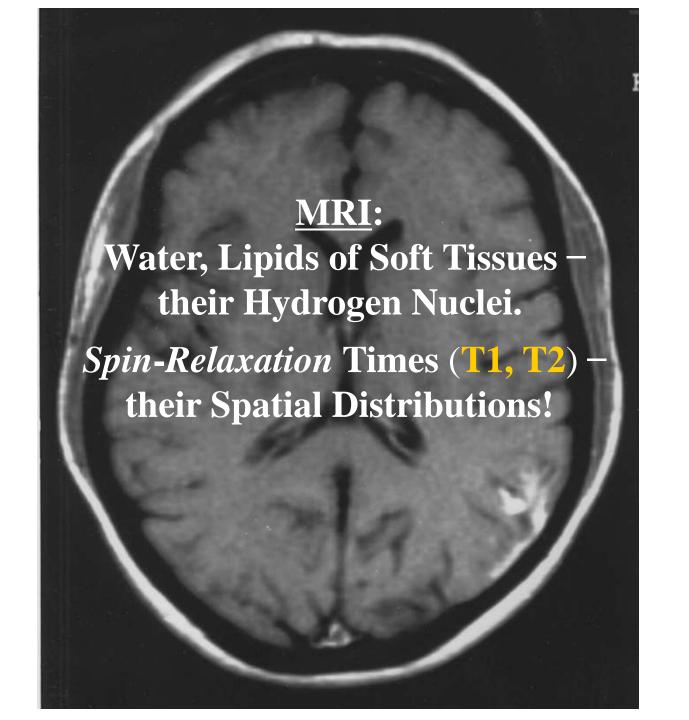
Slow motion ← Rapid

### Average Water T1 Determined by $J(f_{Larmor})$



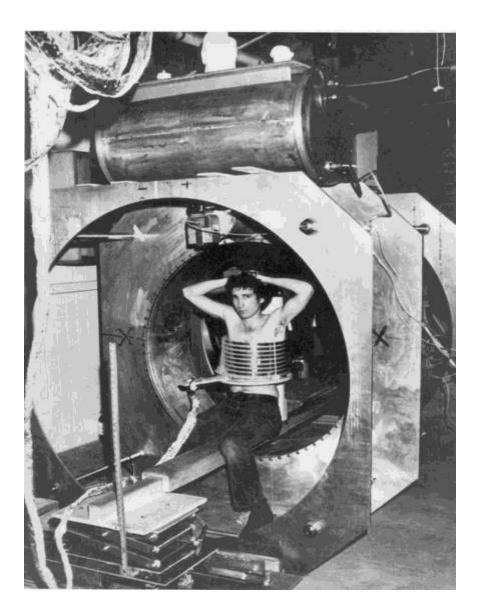
Magnetic noise (water tumbling, *etc.*) frequency

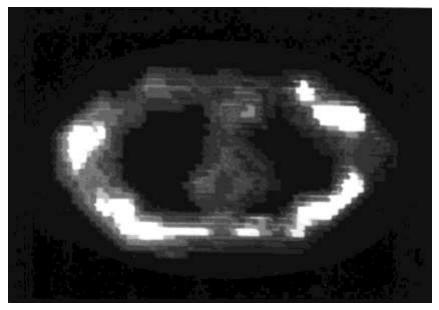
Slow motion ← Rapid



### Sketch of the MRI Device

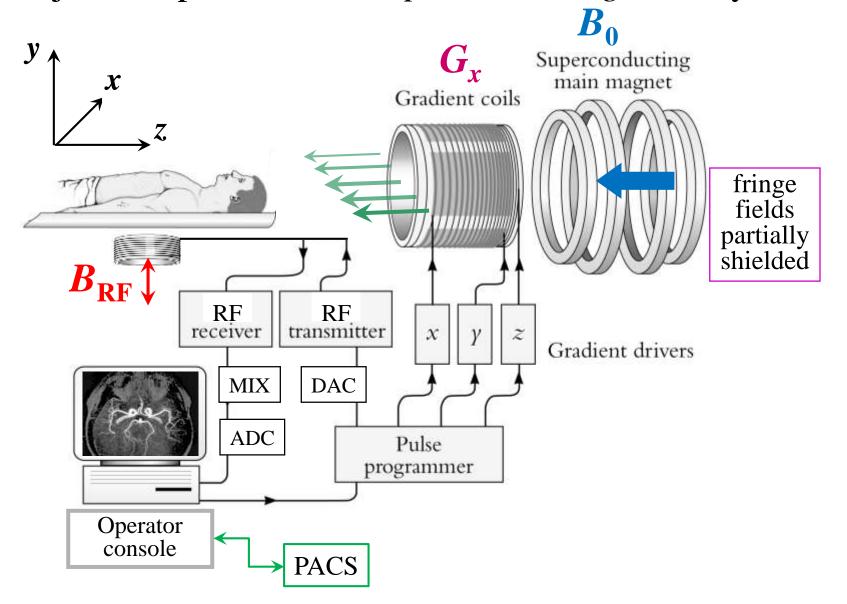
### Damadian's Indominable, 1977 (Smithsonian)



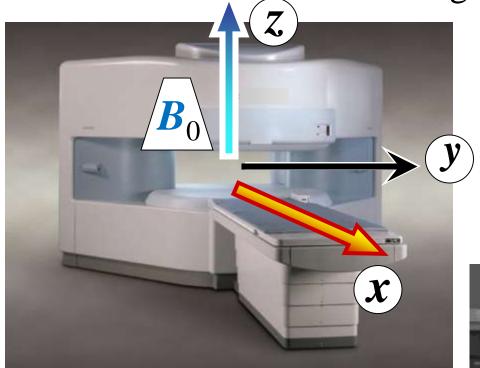


Nobel Prize, 2003
Paul Lauterbur
Peter Mansfield

### Major Components of a Superconducting MRI System



**MRI** Magnets



Open (most of this presentation) electromagnetic or permanent

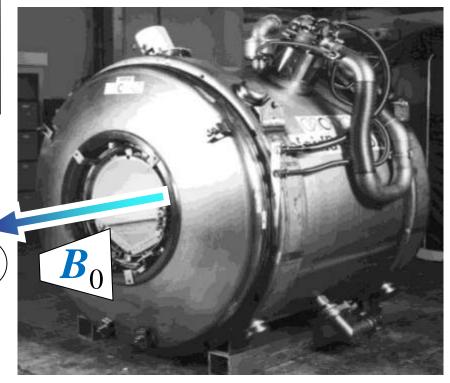
Superconducting *e.g.*, niobium-titanium

 $B_0$ : < 0.5T, 1.5 T, 3.0 T, (7 T) Homogeneity: 10 ppm

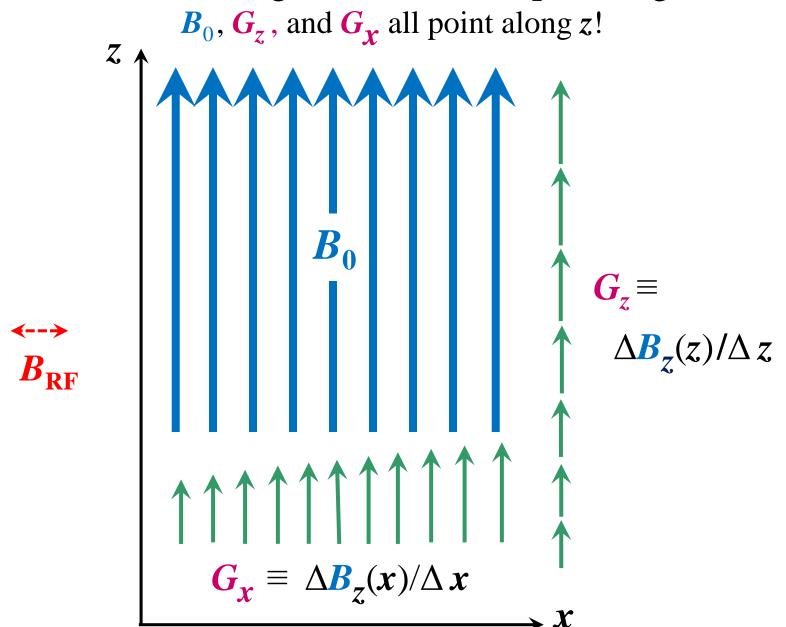
Shielding: passive and active

Cryogen: 0.1 liter He/y

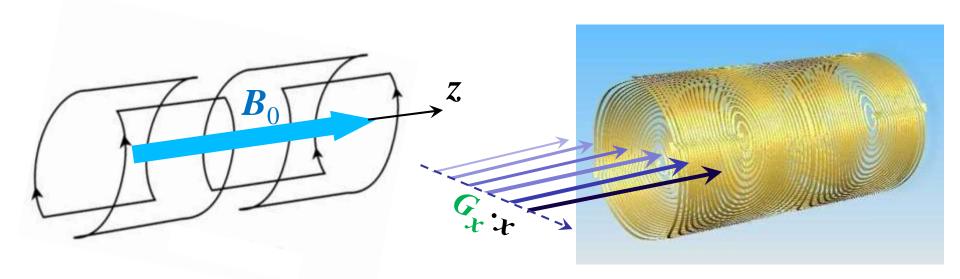
Weight: 4 tons (supercond.)



### Three External Magnetic Fields in *Open Magnet MRI*



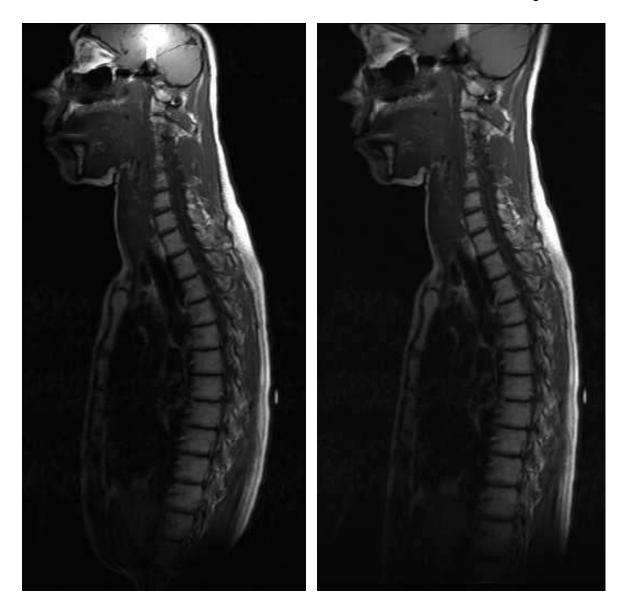
#### x-Gradient Magnet Winding for Superconducting Magnet



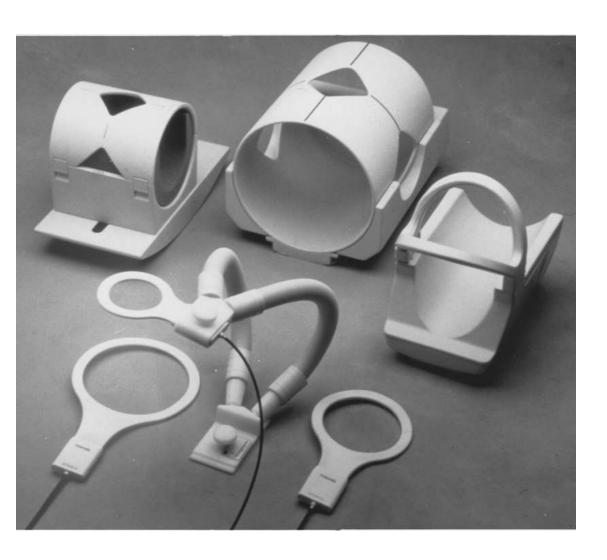
one layer of x-gradient coil

x-Gradient  $dB_z/dx$  20 - 60 mT/ m Rise time 0.3 ms (to reach 10 mT/ m) Slew rate 50 - 200 mT/m/ms

#### Artifact: Gradient Non-Linearity



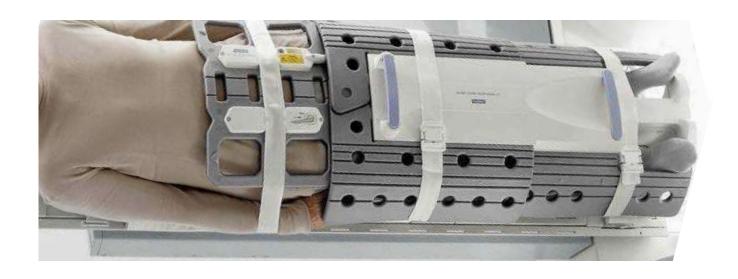
#### RF Coils



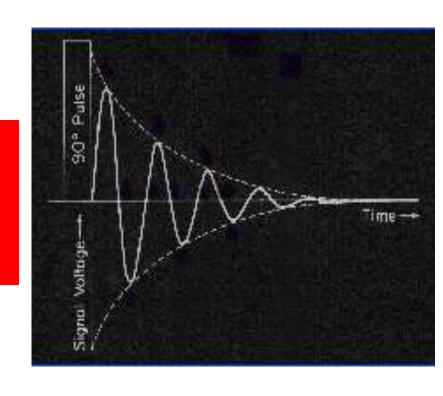


 $B_{\rm RF}$ : 20  $\mu T$ Pulse on-time: 3 msec RF power:  $15-25~{\rm kW}$ SAR:  $2-20~{\rm W/kg}$ 

# 'Parallel' RF Receiving Coils for Much Faster Imaging transmit coils coming



# Part 3: "Classical" MRI



'Classical' Approach to NMR FID Image Reconstruction, *k*-Space

### 'Classical' Approach to NMR; FID Image Reconstruction, *k*-Space

Normal modes, resonance, precession, nutation Free Induction Decay (FID) in a voxel, without the decay *e.g.*, very slow T1 relaxation

Untangling the FID RF signals from a row of voxels:

Temporal Fourier series approach

FID imaging of the two-voxel 1D patient

FID Imaging via k-Space; Spatial FT

#### The Two Approaches to NMR/MRI (incompatible!)

quantum state function

|ψ⟩ <u>now this</u>

Simple QM  $|\uparrow\rangle$ ,  $|\downarrow\rangle$ 

transitions between spin-up, spin-down states

 $f_{\text{Larmor}}$ ,  $m_0$ , T1

oversimplified

Classical Bloch Eqs. for expectation values

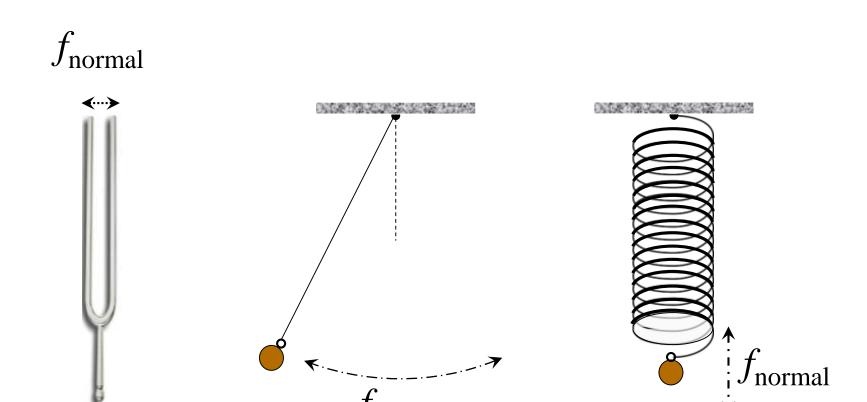
precession, nutation of voxel magnetization, m(t)

 $f_{\text{Larmor}}$ , T2, k-space

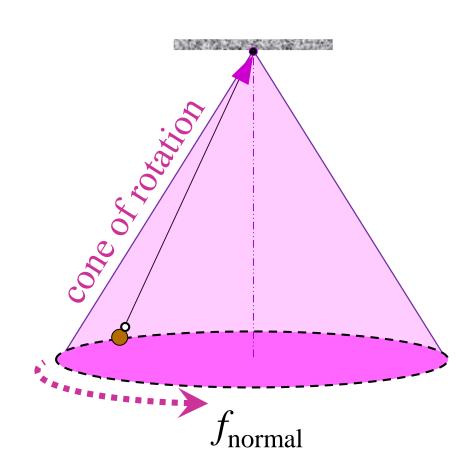
exact; from full QM



#### Normal Mode, at $f_{\text{normal}}$



#### A Normal Mode of a 2-D Pendulum



#### Normal Mode *Precession* about External Gravitational Field

J(t): Angular Momentum

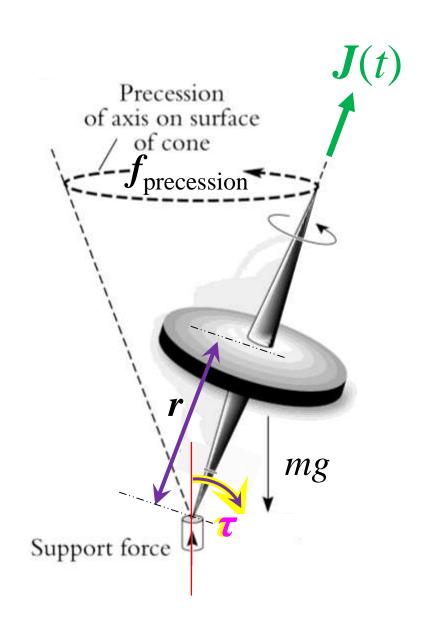
With torque,  $\tau$ :

$$dJ/dt = \tau$$

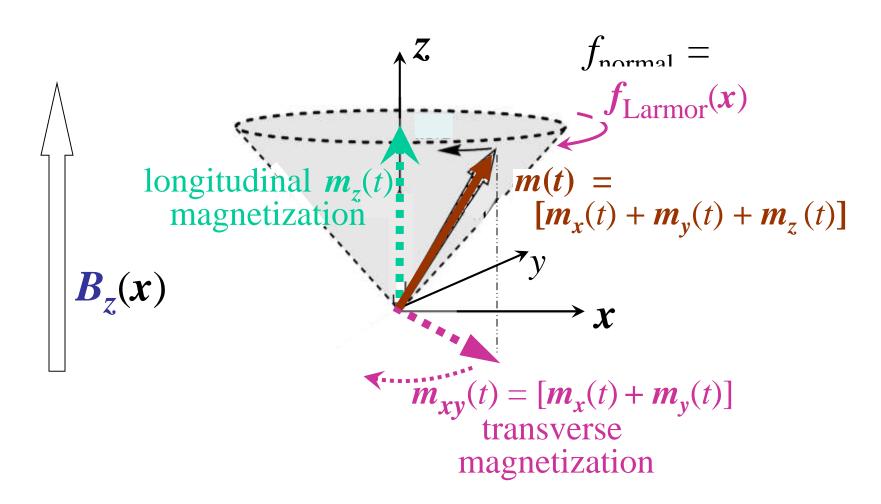
**Equation of Motion** 

Angular acceleration is just like dp / dt = F but here, J(t) changes <u>direction</u> <u>only:</u>

**Precession** at  $f_{\text{normal}}$ 



## Normal Mode Precession of Voxel m(x,t) in Magnetic Field can be derived rigorously from quantum mechanics



#### Classical Bloch Equations of Motion for m(x,t) in $B_0$

Equation of motion for spinning body

$$dJ/dt = \tau$$
 (external torque)
$$d(\mu/\gamma)/dt = \tau$$
 recall:  $\mu = \gamma J$ 

Lorentz torque on spins at x with magnetic moment  $\mu$  in  $B_z(x)$ :

$$\tau = \mu \times B_z(x)$$
 (vector cross product)

Equation of motion becomes:

$$d\mu(x,t)/dt = \gamma \mu(x,t) \times B_{z}(x) .$$

Sum/average over all protons in voxel:

$$d\langle m(x,t)\rangle/dt = \gamma\langle m(x,t)\rangle \times B_{z}(x)$$

With T1 relaxation along z-axis:

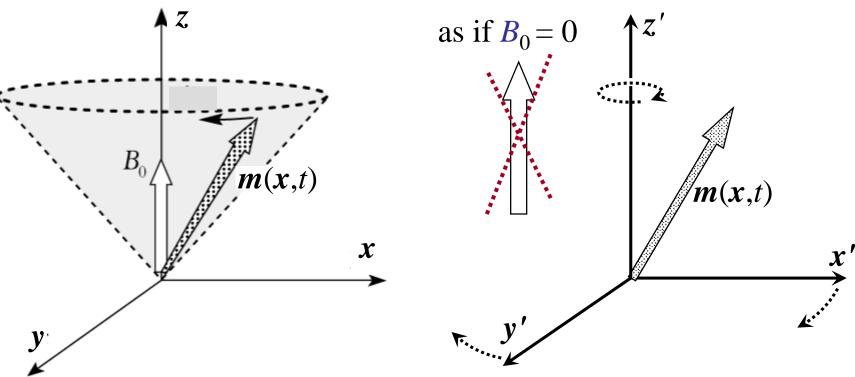
$$d\langle m(x,t)\rangle/dt = \gamma \langle m(x,t)\rangle \times B_z(x) + [\langle m(x,t)\rangle - m_0(x)]\hat{z} / T1$$

Expectation Value,  $\langle m(x,t) \rangle$ , behaves *classically* 

Precession of m(x,t) about  $B_0$  at  $f_{Larmor} = (\gamma/2\pi) B_z(x)$ as seen from a frame that is

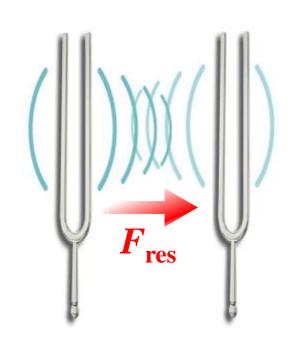


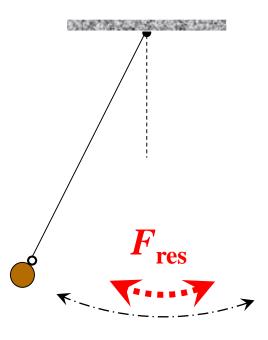


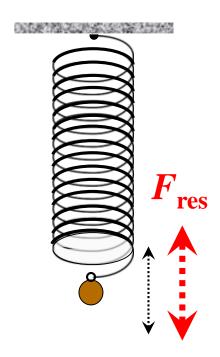


The ponies don't advance when you're *on* the carousel; It's as if  $B_0 = 0$ !

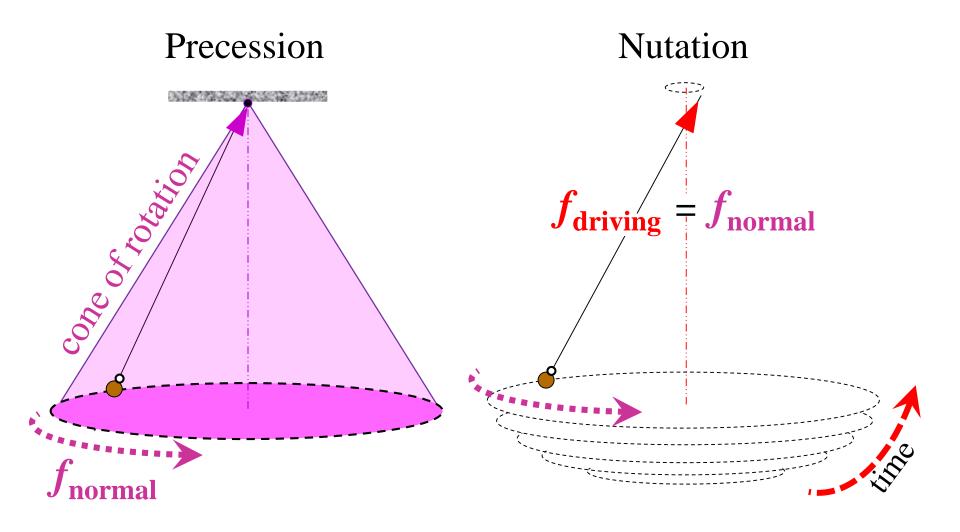
#### Resonance Energy Transfer when $f_{\text{driving}} = f_{\text{normal}}$



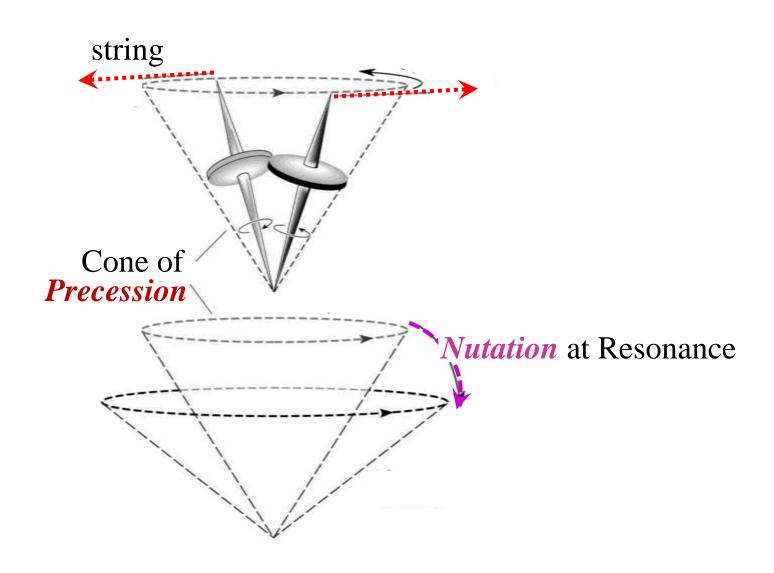




#### Normal Mode and Resonance of a 2-D Pendulum

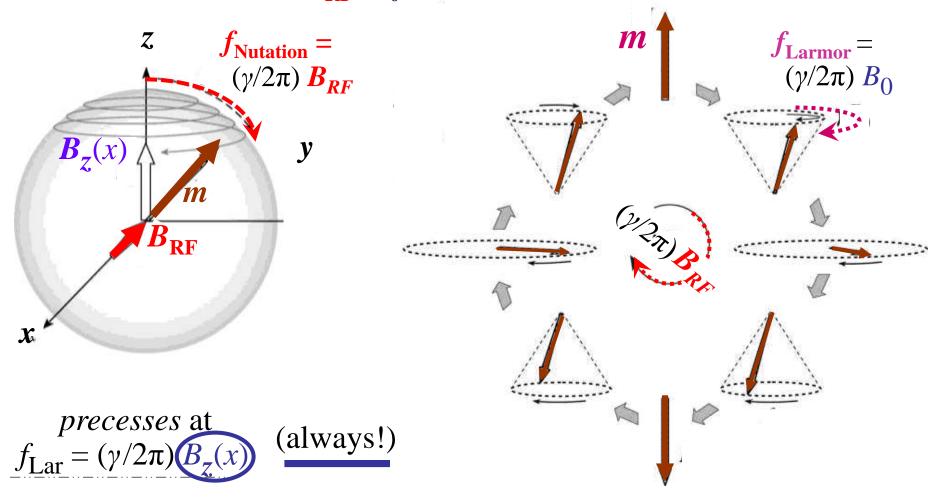


#### And Now for Something Completely Different: Nutation



#### <u>Nutation</u> of the Voxel's Magnetization, m(x,t)

$$B_{RF}/B_0 \sim 10^{-5} - 10^{-4}$$

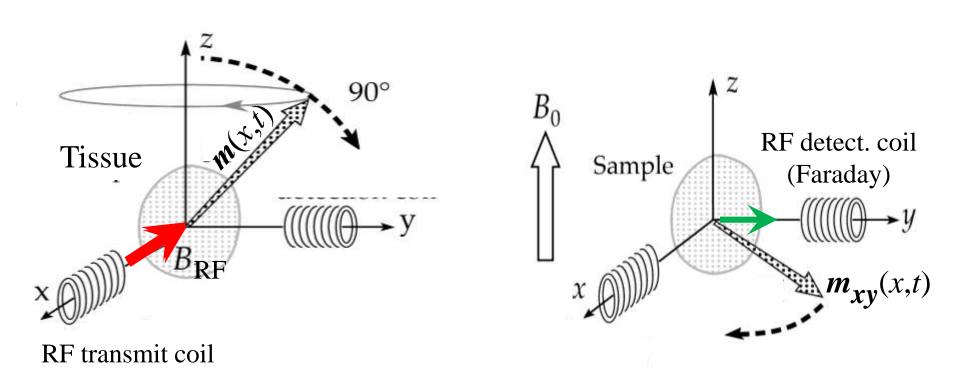


 $nutates at f_{\text{nut}} = (\gamma/2\pi) B_{\text{RF}}$ 

(only when  $B_{RF}$  is on!)

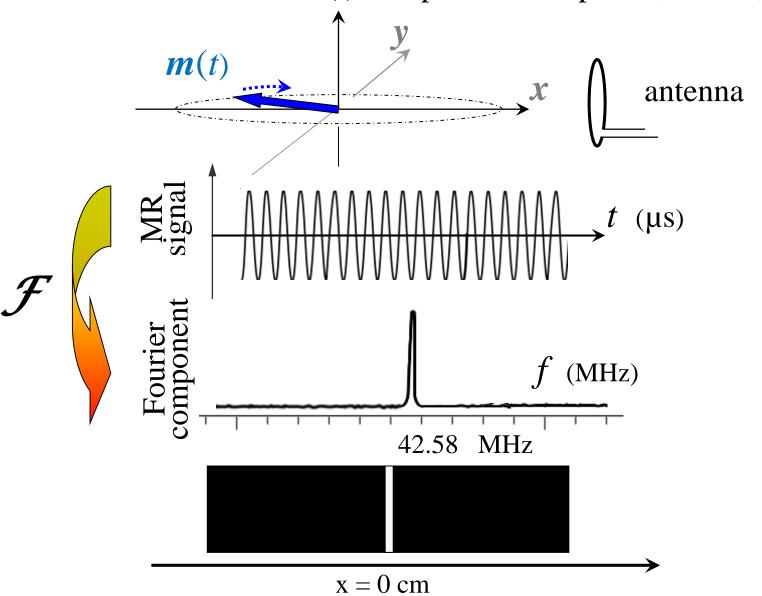
# Free Induction Decay in a voxel, without the decay *e.g.*, very slow T1 relaxation

### m(x,t) for a Single Voxel at x and Precessing in the x-y Plane following a 90° pulse



#### FID Precession, Reception, Fourier Analysis (single voxel)

n.b. measure induced V(t), not power absorption (as before)

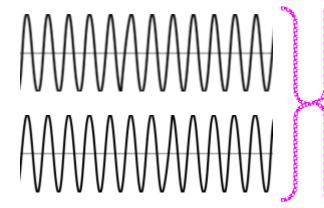


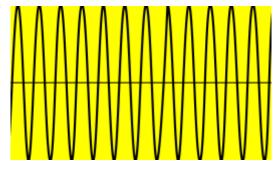
In MRI, the <u>only</u> signal you <u>ever</u> see comes from the set  $\{m(x,t)\}$  all precessing in the x-y plane



# Untangling the FID RF Signal from Two Voxels: temporal Fourier series approach

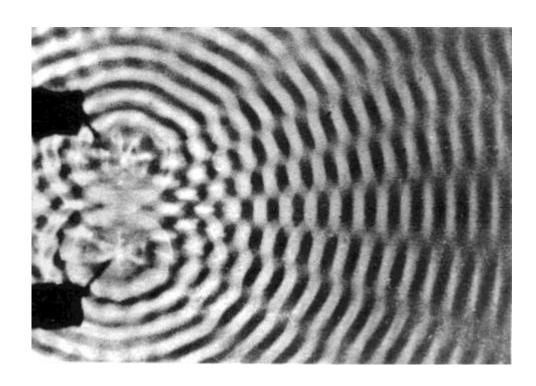
#### Wave Interference in Time or Space



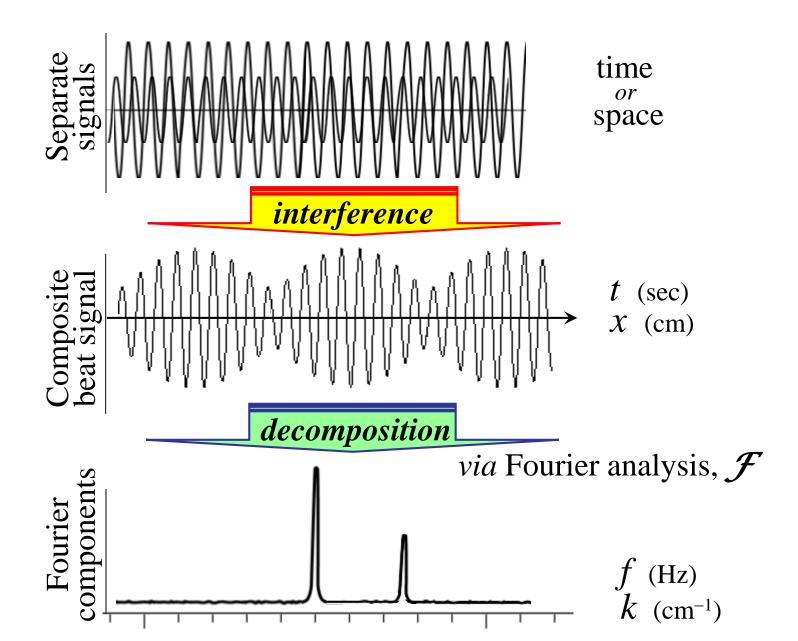


positive (in-phase) interference

negative interference



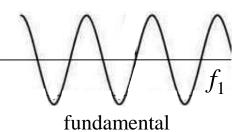
#### Wave Interference and Spectral Analysis



#### Fourier Decomposition of Periodic Temporal Signal

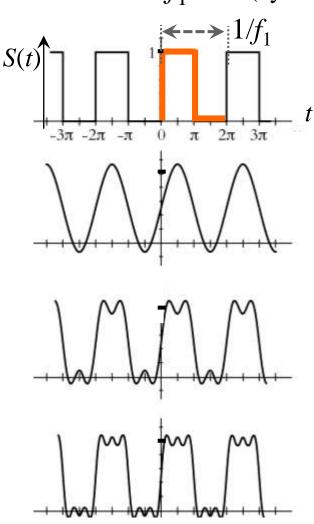
$$S(t) \sim \frac{1}{2} + (2/\pi) \{ \sin(2\pi f_1 t) + \frac{1}{3} \sin(6\pi f_1 t) + \frac{1}{5} \sin(10\pi f_1 t) + \dots \}$$
  
fundamental:  $f_1$  Hz (cycles/sec)

### orthonormal basis vectors

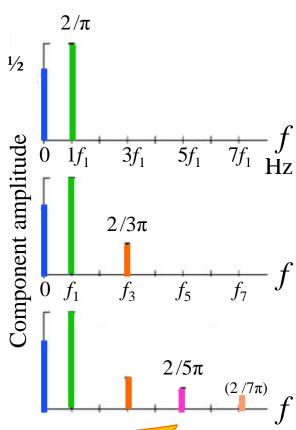


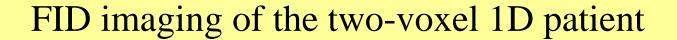


$$f_5$$



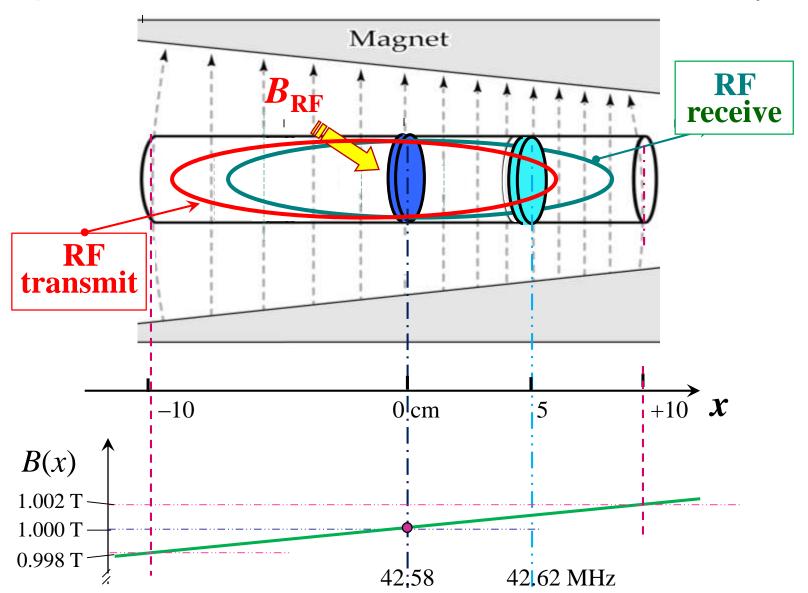
### <u>spectrum</u>





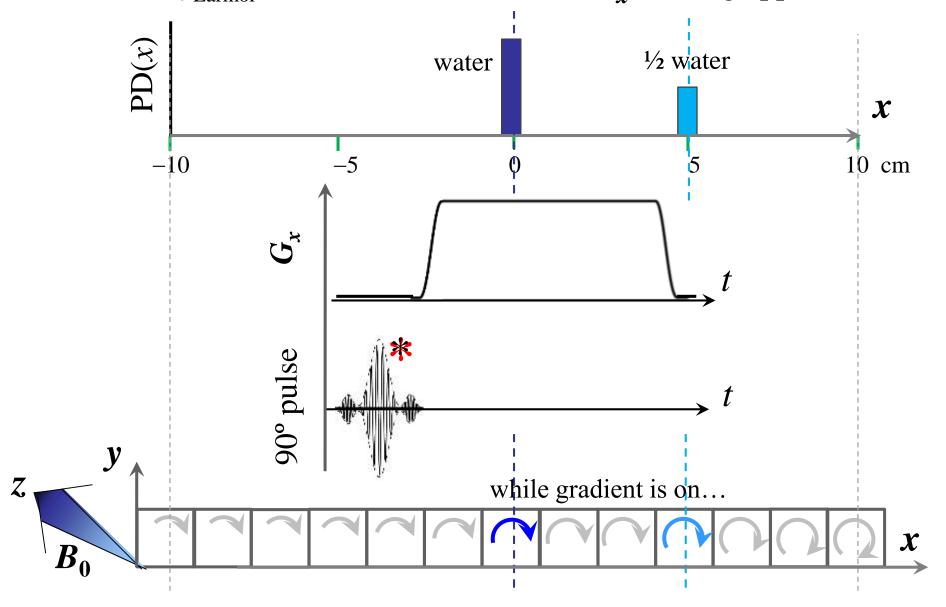
#### $f_{\text{Larmor}}(x) = (\gamma/2\pi) (x \times G_x)$ (in rotating frame)

e.g., FID from 2 water slices, at x = 0 and 5 cm; little decay!

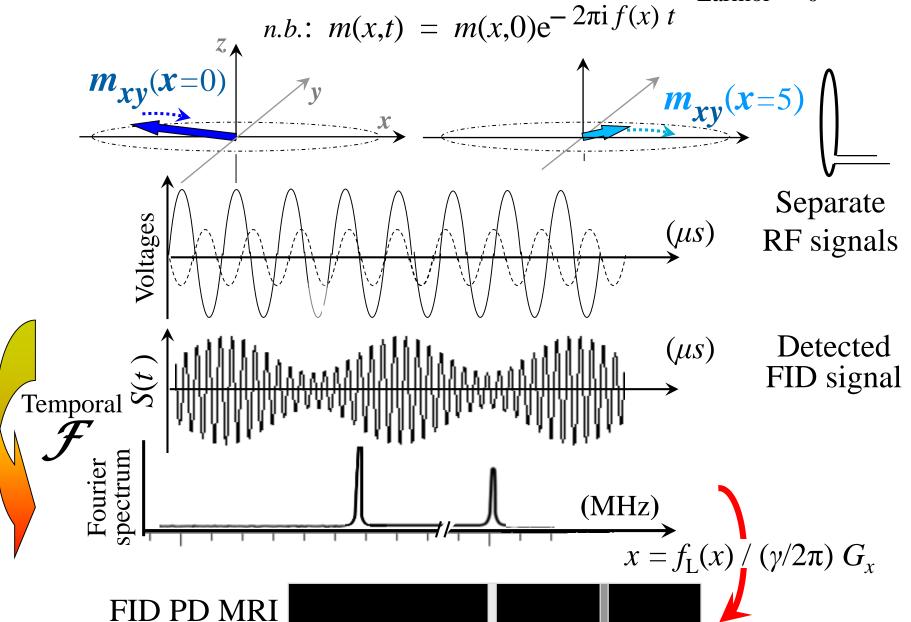


### FID Following Wide-Bandwidth 90° RF Pulse

covers  $f_{Larmor}(x)$  for all N voxels while  $G_r$  is being applied!

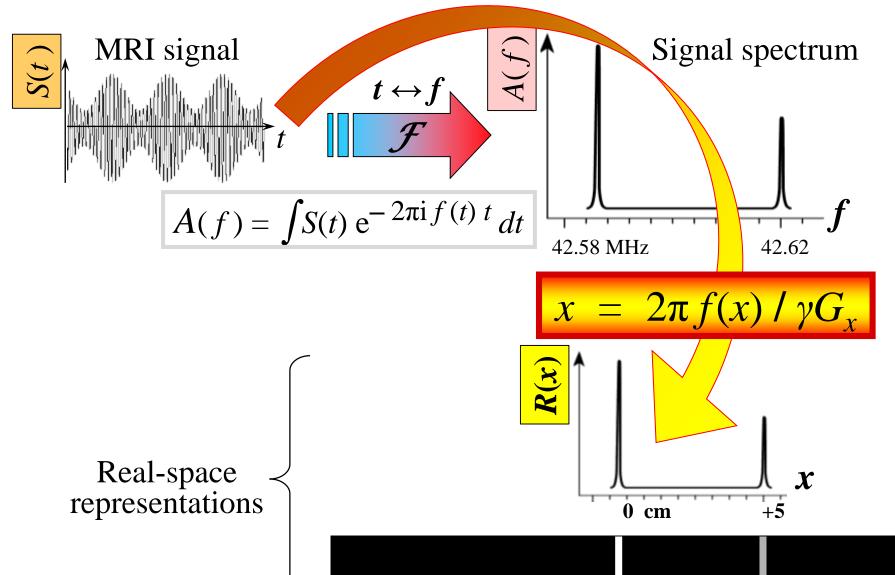


#### FID: Wide-Band RF Pulse, Centered on $f_{Larmor}(B_0)$



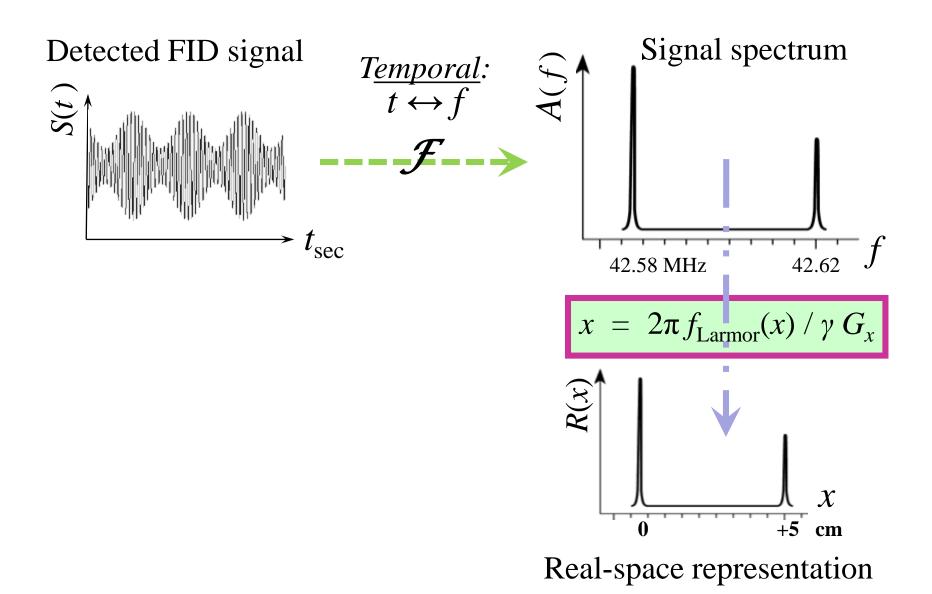
### To Summarize What We Have Done So Far with FID

follow temporal  ${\mathcal F}$  with isomorphism of A(f) to real-space,  ${\it R}(x)$ 



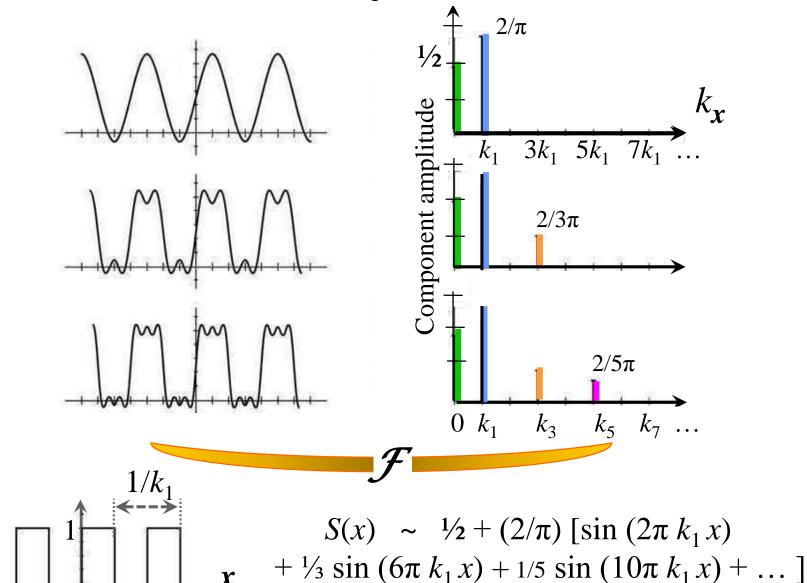
FID Imaging via k-Space; Spatial FT

#### Again, What We Have Done So Far:



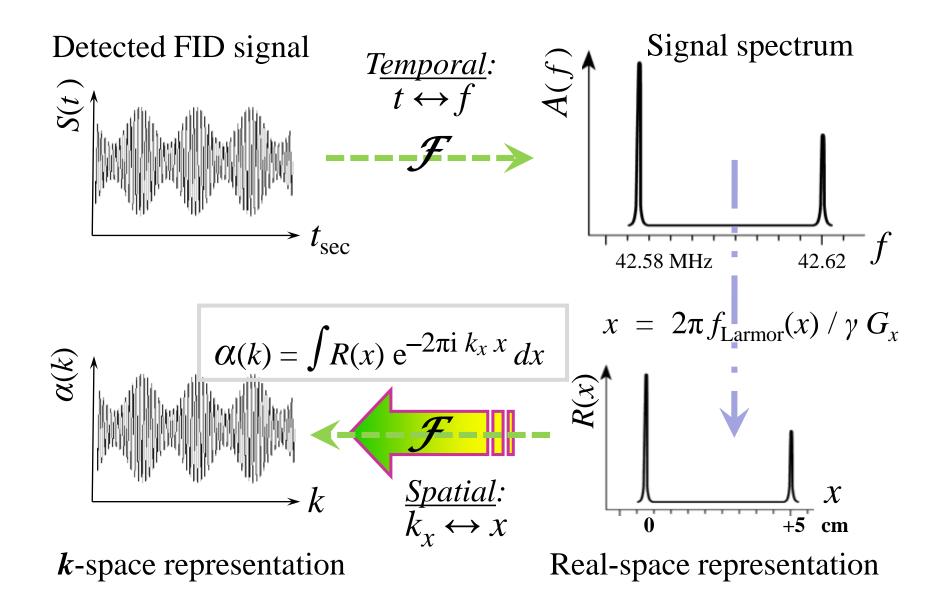
#### Now Consider 1D Spatial Fourier Analysis

fundamental:  $k_1$  (cycles/mm)

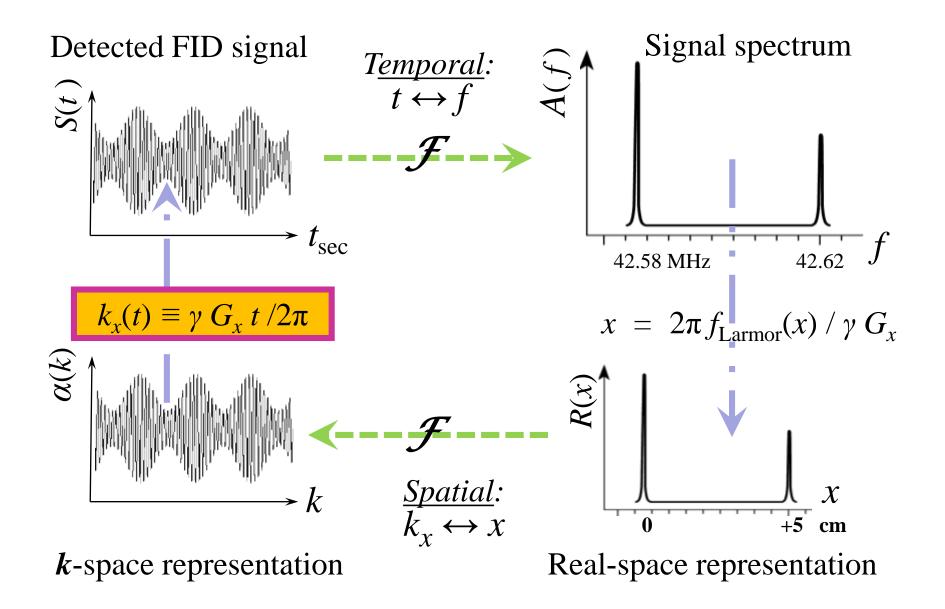


S(x)

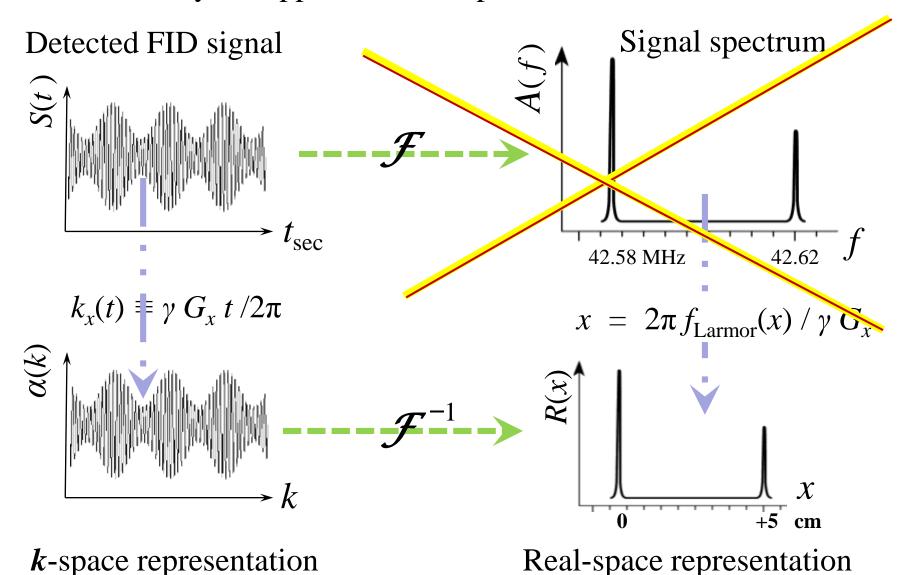
#### New Approach: Continue on with a FT from x- to $k_x$ -Space



#### And Complete the Ring Linking t-, f-, x- and $k_x$ -Spaces

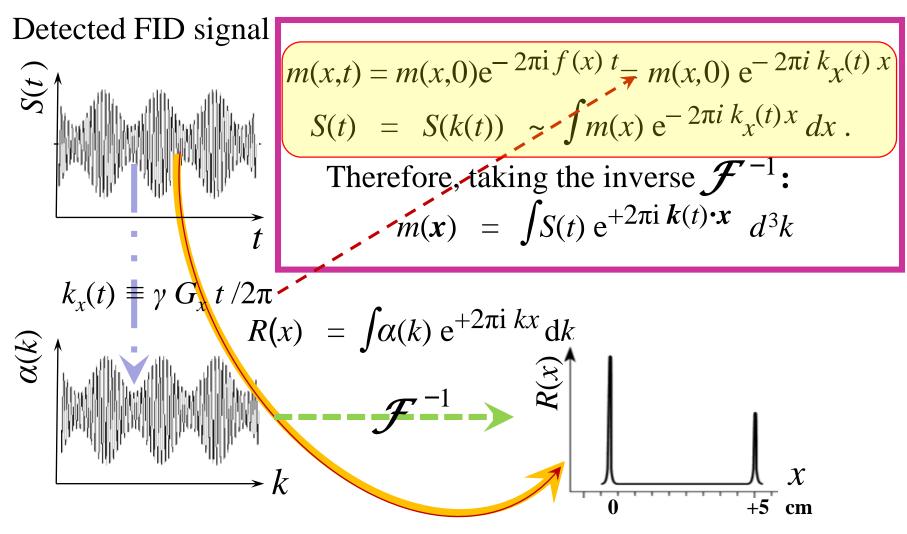


# So, Can Get from S(t) to R(x) in Either of Two Ways! but only the approach $via \, k$ -space works in 2D and 3D



#### k-Space Procedure for FID in 1D Phantom

S(t) comprised of contributions from the m(x) for all x

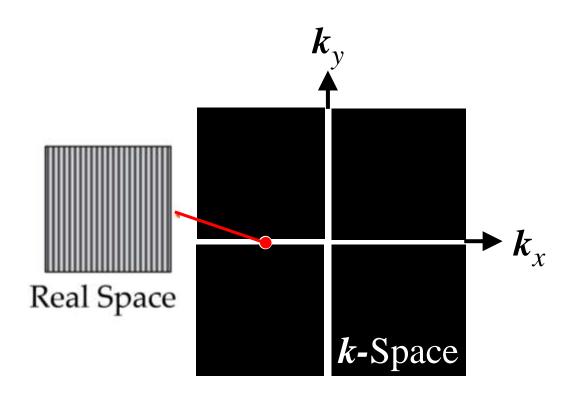


**k**-space representation

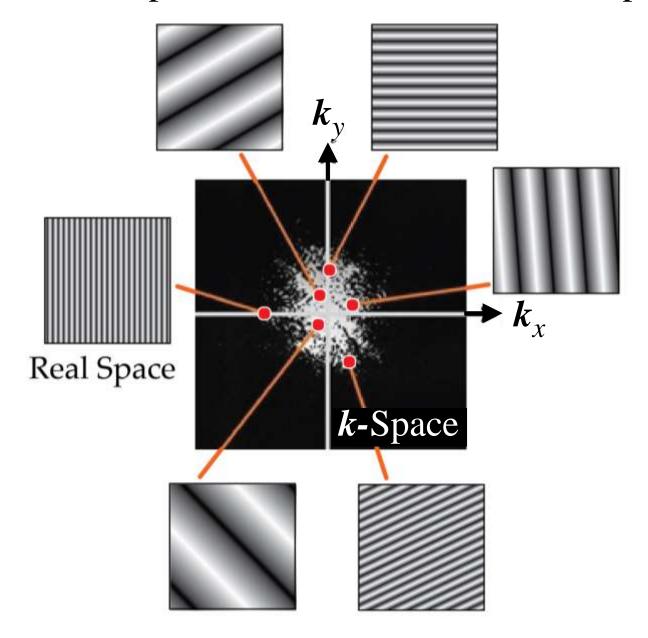
Real-space representation

Spatial Waves, and Points in k-Space

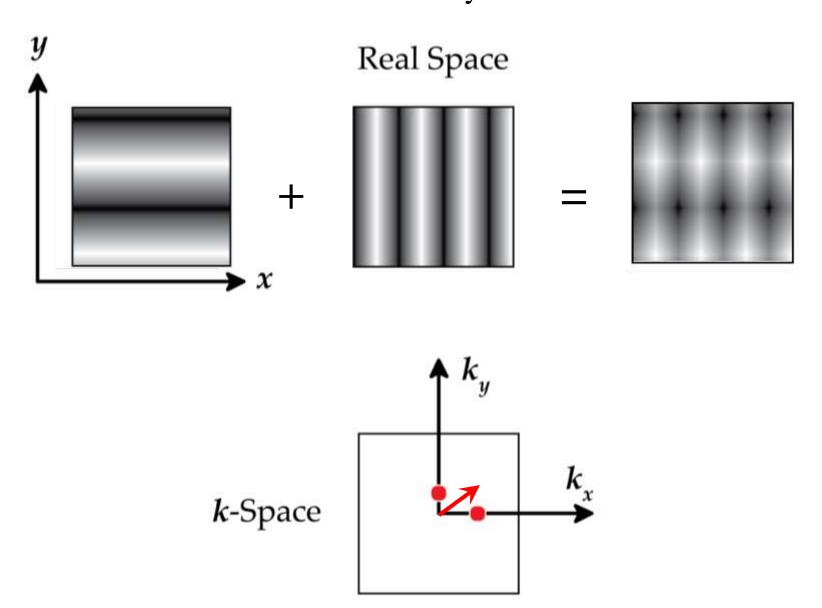
### Point in *k*-Space and Its Associated Wave in Real Space



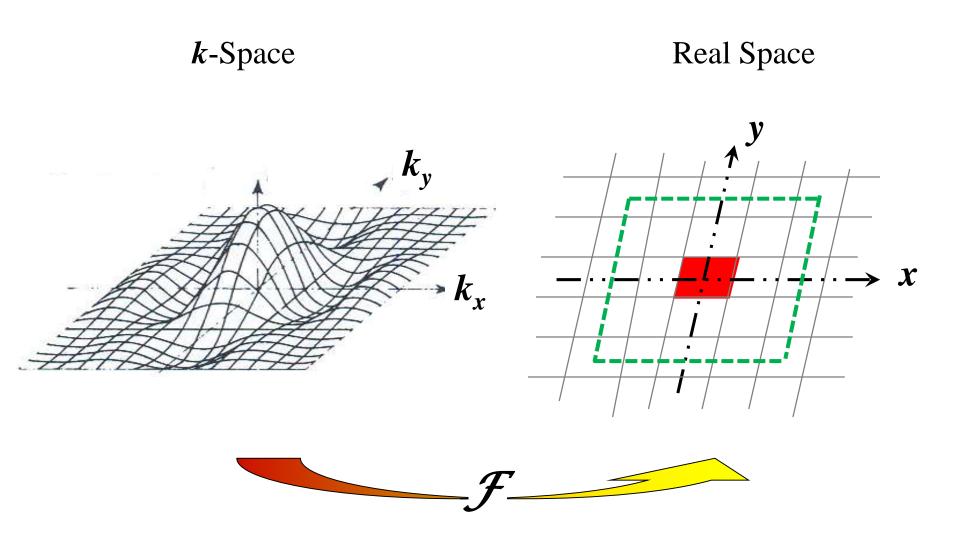
## Points in k-Space and Their Waves in Real Space



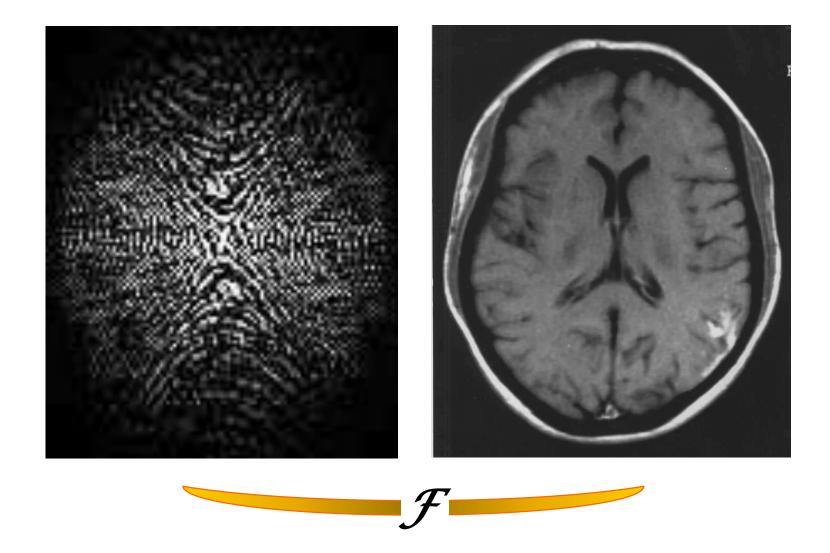
# Composite Patterns in *k*-Space and Real Space linearity



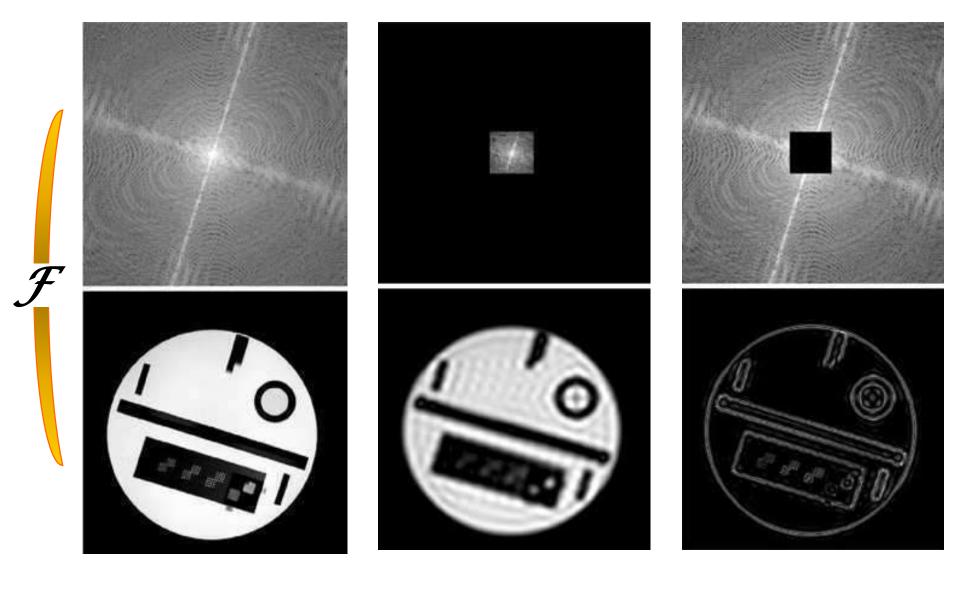
## Fourier Transform of k-Space Pattern to Real Space



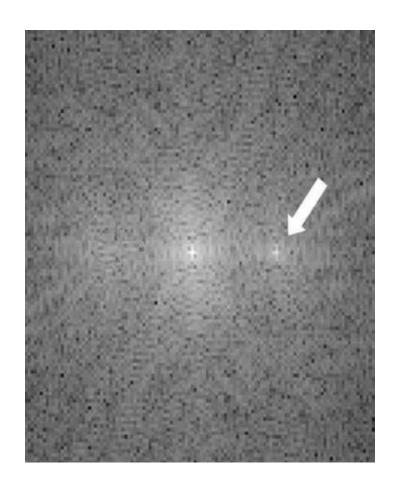
## Fourier Transform of MRI Data in k-Space to Real Space

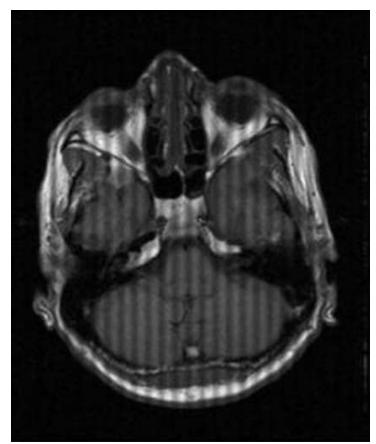


## Fourier Transform from Parts of k-Space to Real Space

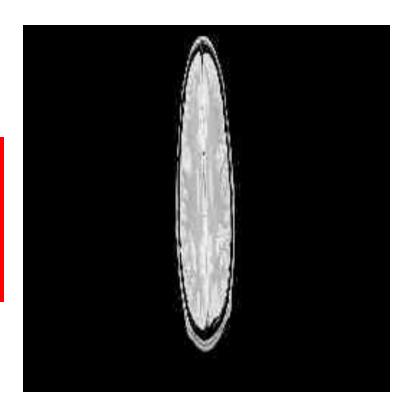


# Herringbone Artifact noise spike during data acquisition





# Part 4: Spin-Warp; 2D

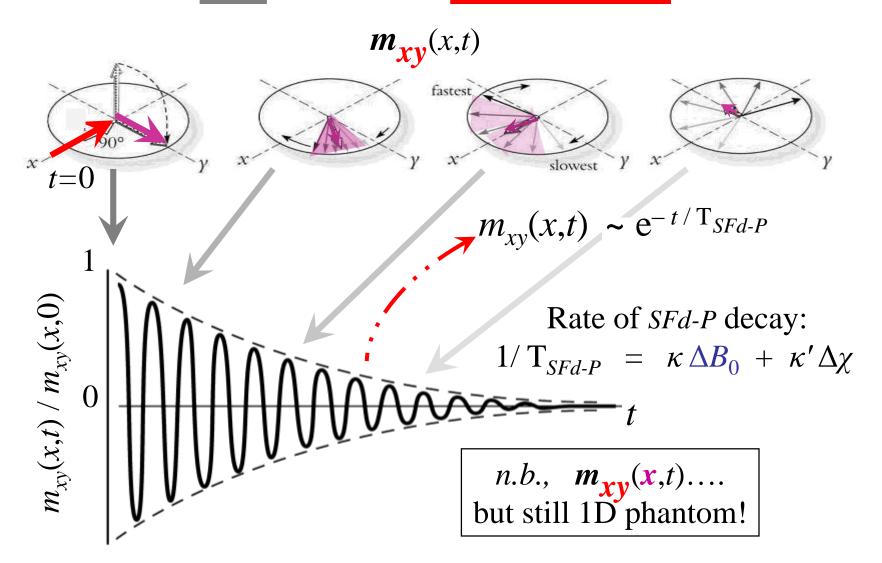


Spin-Echo Reconstruction
T2 Spin-Relaxation
T1-w, T2-w, and PD-w S-E MRI
Spin-Echo / Spin-Warp in 2D

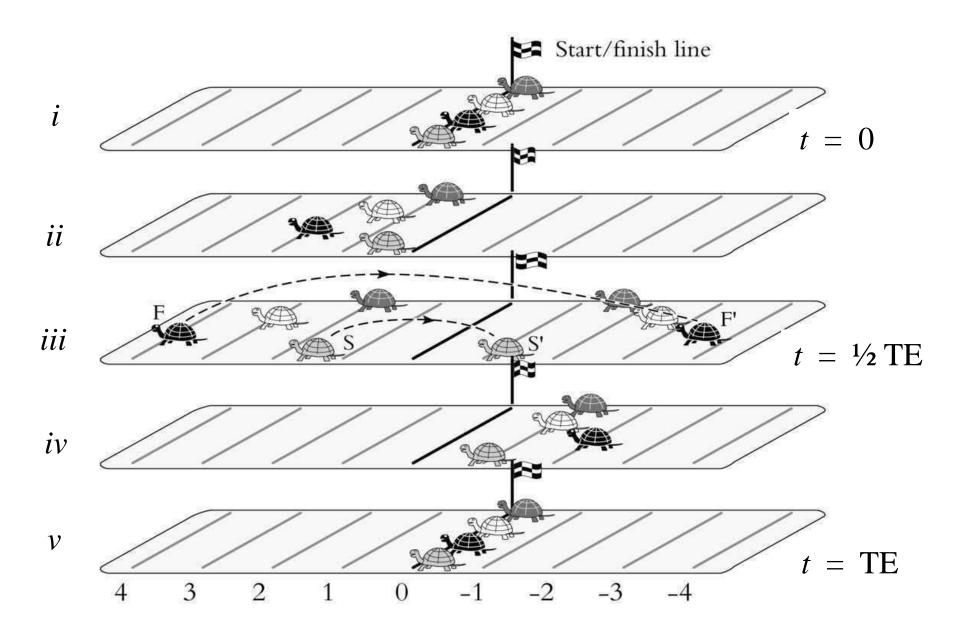
# Spin-Echo Reconstruction

# Static Field de-Phasing in x-y Plane of Spins in a Voxel

<u>SFd-P</u> from <u>static</u>-field inhomogeneities & susceptibility effects
<u>This</u> de-phasing is <u>REVERSABLE!!!!</u>



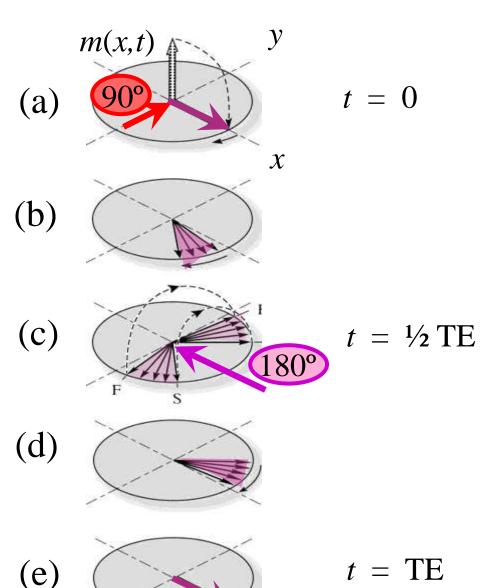
## Kentucky Turtle-Teleportation Derby



### Spin-Echo Generates an Echo and Eliminates SFd-P Effects

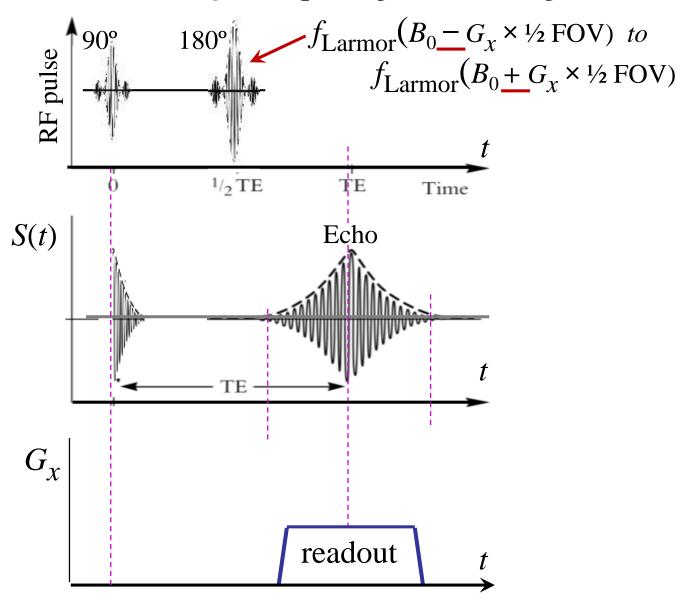
Local magnetic environment for each spin packet is *static*. So *this* de-phasing can be REVERSED!!!

(**T2** dephasing, by contrast, is **random** over time, and can *NOT* be un-done!!)



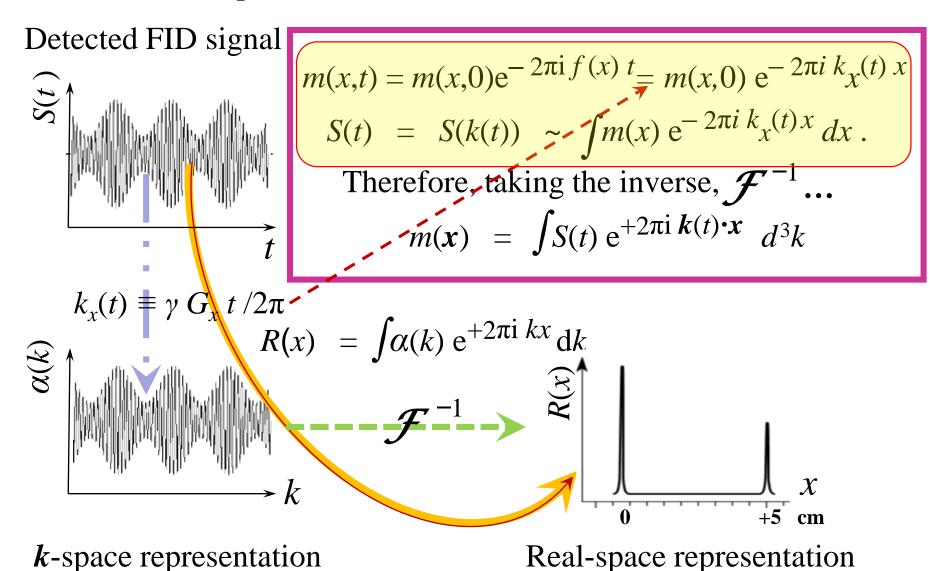
#### S-E Pulse Sequence

with 256 voxels, e.g., sample/digitize echo signal 512 times



### Again: k-Space Procedure for FID in 1D

S(t) comprised of contributions from the m(x) for all x



## During Readout, $k_x$ Increases Linearly with t

Signal is sampled sequentially at 512 times,  $t_n$ , after  $G_x$  is turned on.

k(t) for all voxels increases linearly with t while the echo signal is being received and read:  $k_n = [G_x \gamma / 2\pi] t_n$ . Later sampling times, hence larger k-values, correspond to greater spatial frequencies!

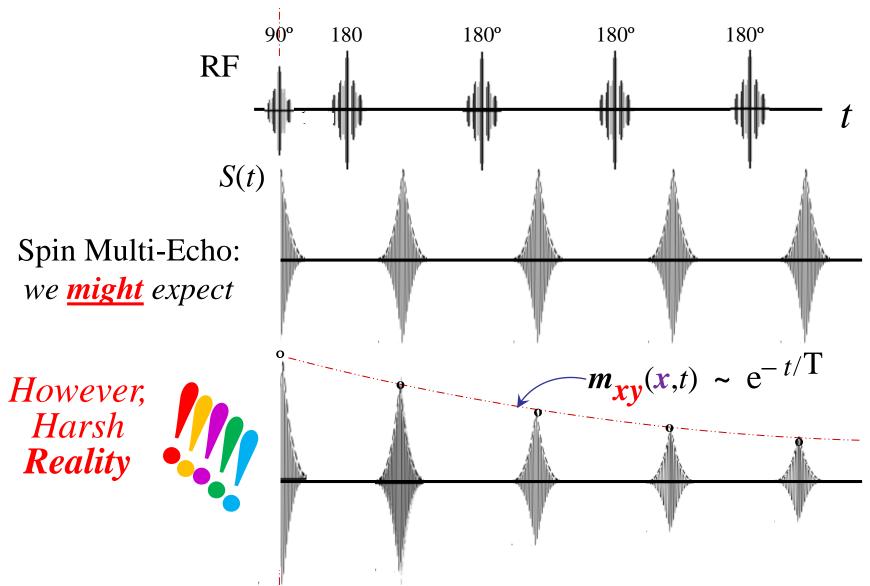
S-E reads out from  $k_n = -k_{max}$  to k = 0 to  $k = k_{max}$   $k = -k_{max} \qquad k = 0 \qquad k_{max}$   $t_n \qquad t_{51}$ 

## T2 Spin-Relaxation

T1-Events
and *Non-Static*, *Random*, *Non-Reversible*Proton-Proton Dipole Interactions *Both* Contribute to 1/T2

Specifically, any process that decreases either the number of transverse spins or their phase coherence will add to T2 relaxation.

# Thus, Spin-Echo Signal *Does* Decay, and Much Faster than 1/T1!



SOMETHING BIG

MUST BE GOING ON

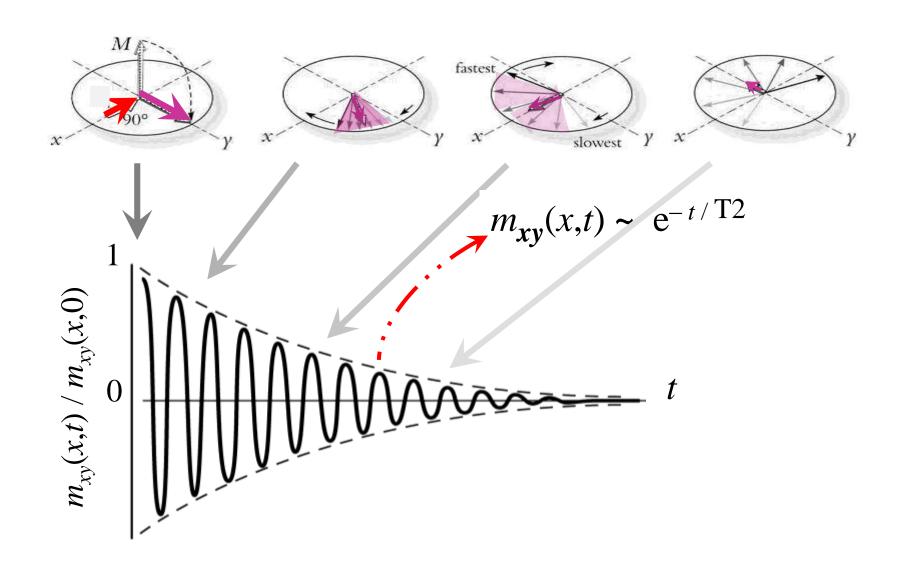
IN ADDITION TO

THE T1 PROCESS

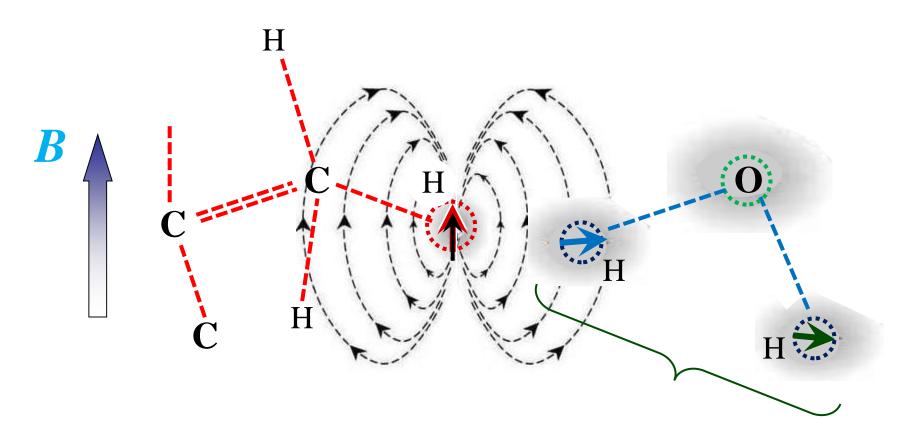
Another Form of Relaxation, As Well!

Call it **T2**!

# Even with S-E, Spin De-Phasing Is Caused by T1 Events and by *Non-Static*, *Random*, Proton-Proton Dipole Interactions



# Spin-Spin (Secular) De-Phasing in Bound *Water* with *its* protons precessing in the *x-y* plane at different rates



Quasi-static *spin-spin interactions* do *not* involve exchange of energy!

## $1/T2 \sim 10 \times (1/T1)$

Tissue	PD p+/mm <sup>3</sup> , rel.	T1, 1T (ms)	T1, 1.5T (ms)	T1, 3T (ms)	T2 (ms)
pure H <sub>2</sub> 0	1	4000		4000	4000
brain					
CSF	0.95	2500	2500	2500	200
white matter	0.6	700	800	850	90
gray matter	0.7	800	900	1300	100
edema			1100		110
glioma		930	1000		110
liver			500		40
hepatoma			1100		85
muscle	0.9	700	900	1800	45
adipose	0.95	240	260		60

Exponential, T2-Caused De-Phasing of  $m_{xy}(x,t)$  in x-y Plane

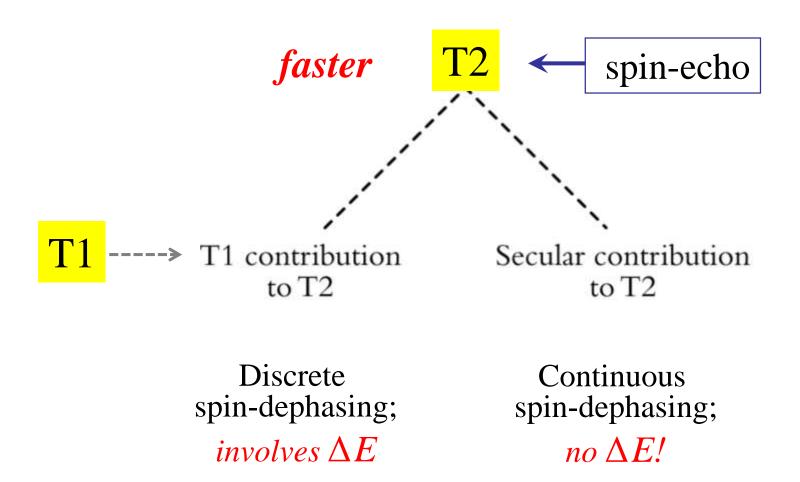
$$0$$

$$m_{xy}(t)/dt = -(1/T2) m_{xy}(t)$$

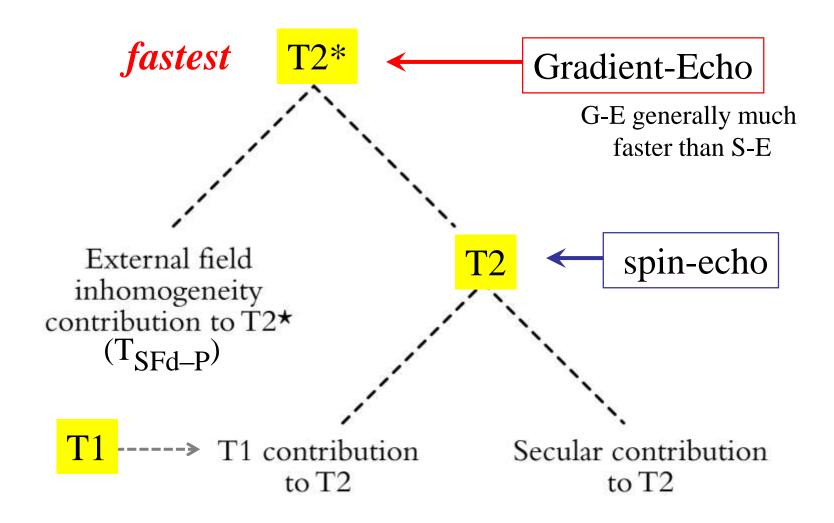
$$m_{xy}(t)/m_{xy}(0) = e^{-t/T2}$$

$$d\mathbf{m}(x,t)/dt = \gamma \mathbf{m}(x,t) \times \mathbf{B}_{\mathbf{z}}(x) - [\underline{m}(x,t) - \underline{m}_{\mathbf{0}}(x)]\hat{\mathbf{z}} - [\underline{m}_{x}\hat{\mathbf{x}} + \underline{m}_{y}\hat{\mathbf{y}}]$$
 classical Bloch Equation T1

$$1/T2 = 1/(2 T1) + 1/T2_{\text{secular}}$$



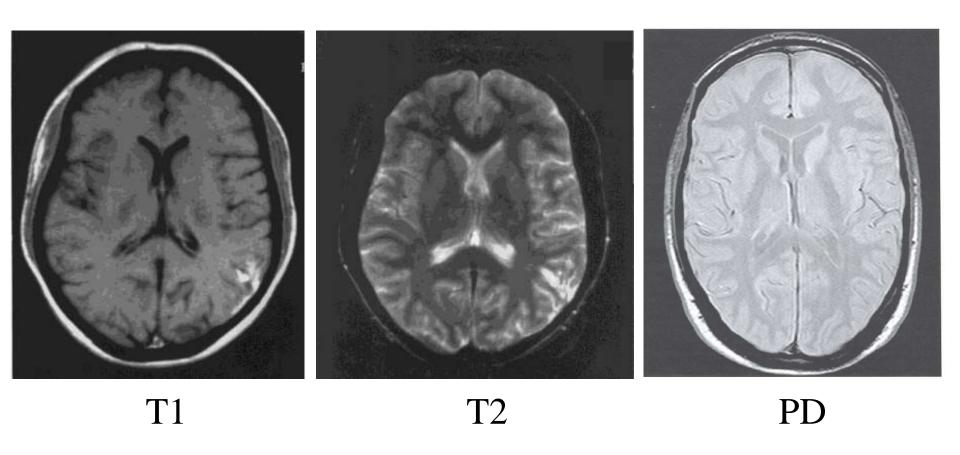
## One Last Member of the Spin-Relaxation Family Tree: T2\*



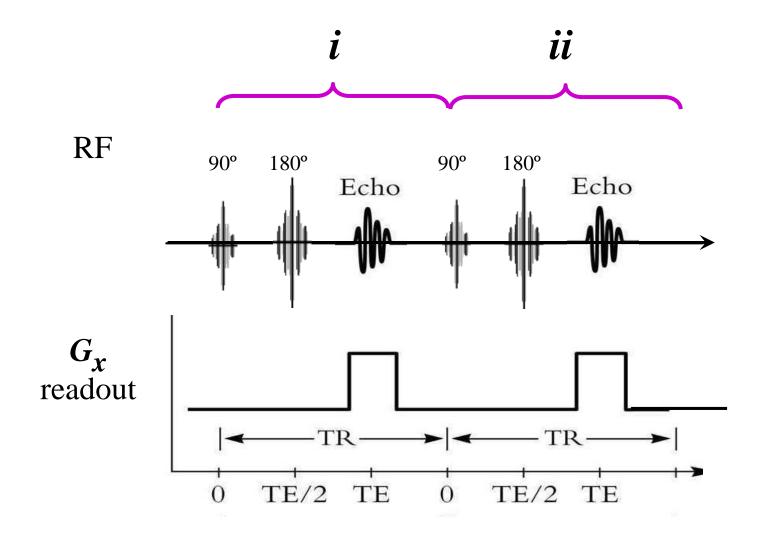
# T1-w, T2-w, and PD-w S-E MRI

#### Three Different Forms of MRI Contrast

created by, and reflecting, three quite different physical processes



## Multiple Spin-Echo Pulse Sequence



### MRI Signal Strength at t = TE Depends on....

Inherent Parameters	Operator Settings	time since	
<b>T</b> 1	TR ←	previous	
T2	TE	S-E	
PD		sequence	

$$S(t = TE)$$
 ~  $PD(1 - e^{-TR/T1}) e^{-TE/T2}$ 

proton prior density regrow

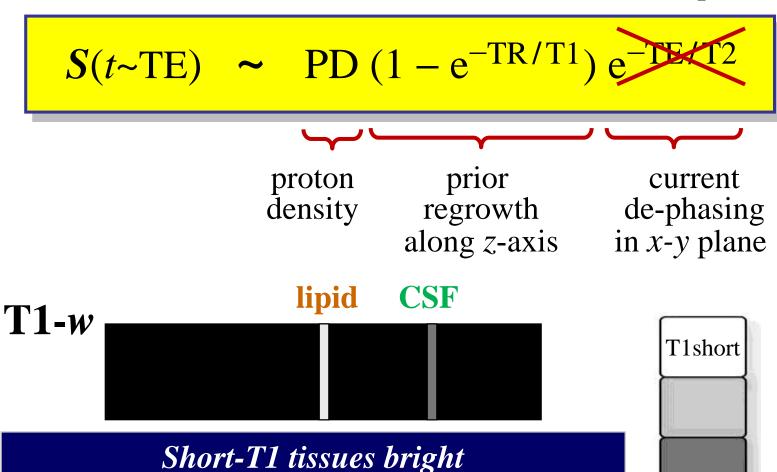
prior current regrowth dephasing along z-axis in x-y plane

bright regions:  $m_{xy}$  large at t = TE

#### T1-weighted — Short TE to Eliminate T2 Contribution

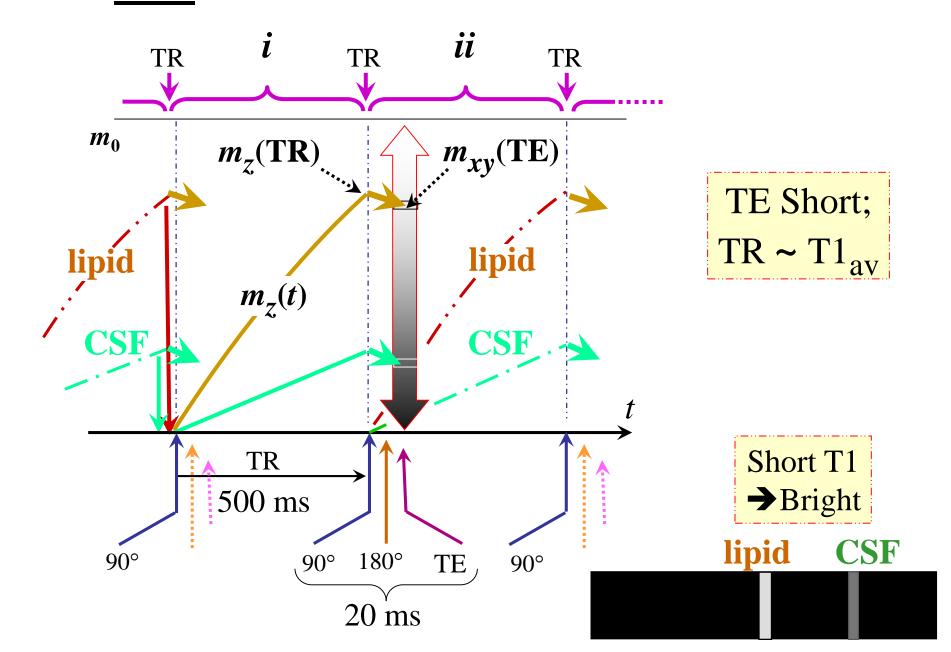
TR ~ T1<sub>av</sub> to maximize contrast

Short TE to minimize T2 impact



(sprightly brightly)

T1-w - Short TE to *Eliminate* T2 Contribution



### <u>T2-w</u> – Long TR to Eliminate T1 Contribution

For T2-w imaging

eliminate T1 impact

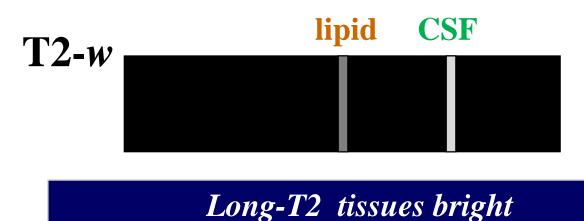
Long TR: TE at mid-T2: maximize contrast

 $S(t \sim TE) \sim PD(1-e)$ 

proton density

prior regrowth along z-axis

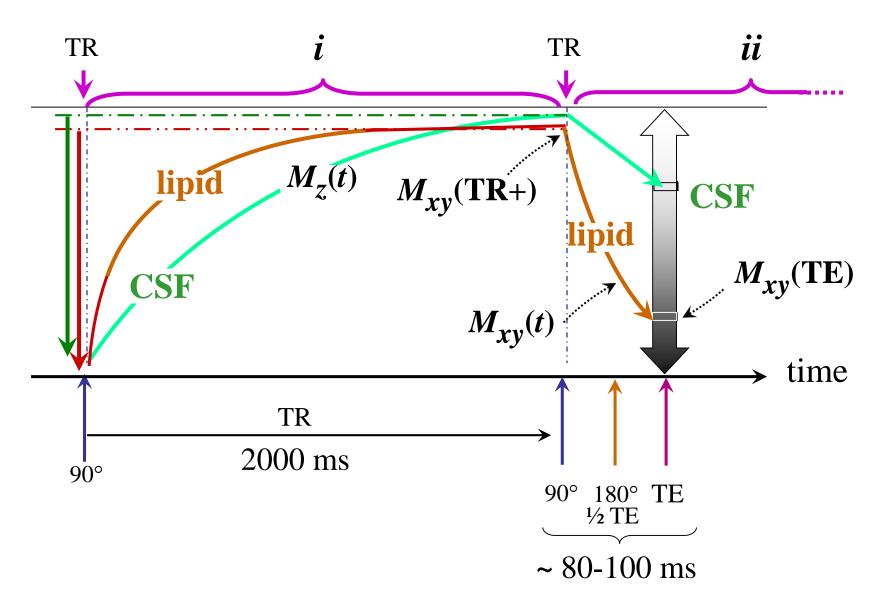
current de-phasing in *x*-*y* plane





### <u>T2-w</u> – Long TR to Eliminate T1 Contribution

TE ~ mid-T2 to maximize T2 contrast

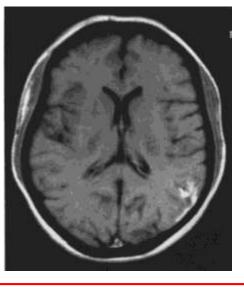


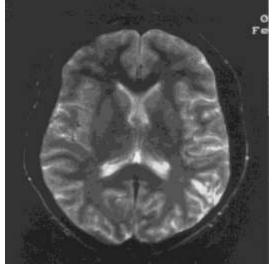
## PD-, T1-, & T2-Weighted Spin-Echo Images

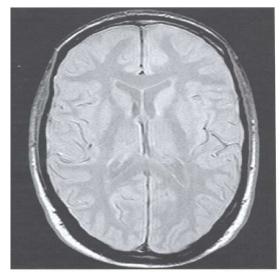
T1-*w* 



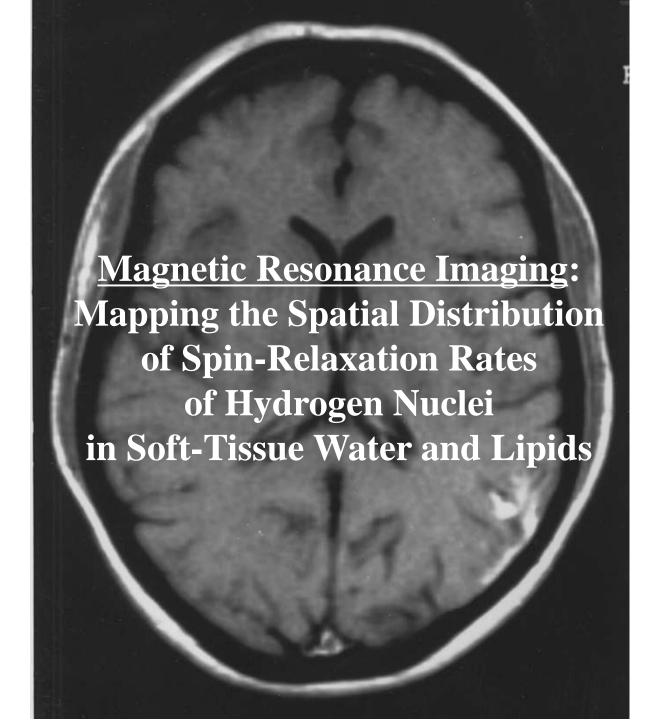








TR	mid- (~T1 <sub>av</sub> )	long	long
(ms)	300 – 700	1,500 – 3,500	1,500 – 3,500
TE (ms)	short	mid- (~T2 <sub>av</sub> )	short
	0 – 25	60 – 150	10 – 25
Bright	short T1	long T2	high PD
SNR	good	lower	best



### Spin-Echo / Spin-Warp in 2D

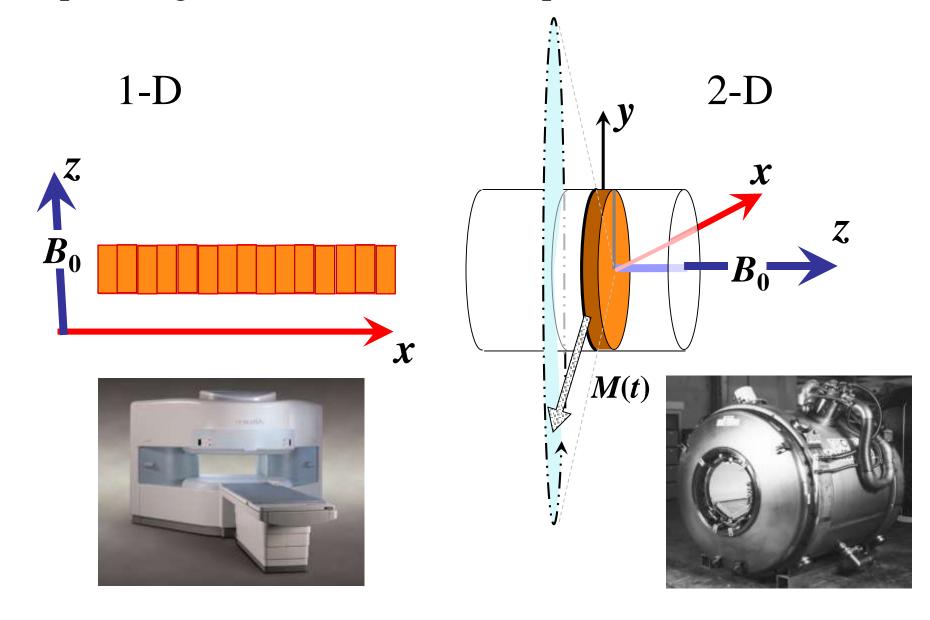
#### Sensitive Point (Earliest) Reconstruction

#### Oscillating Gradient Fields

net field stable in one voxel at a time; move sensitive point around.

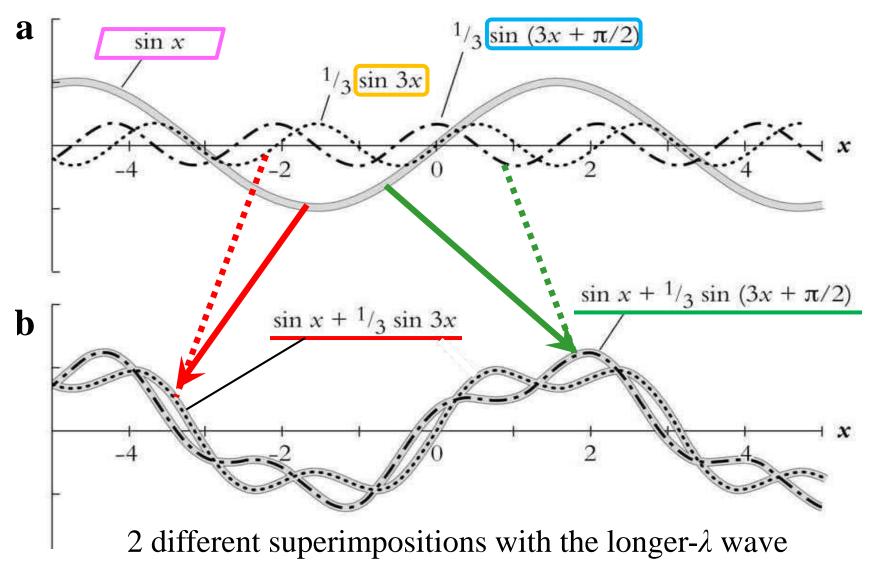
cannot be done with CT.

#### Open Magnet, 1D Phantom vs. Supercon, 2D Phantom



#### In 2-D MRI, Phase Is Every Bit As Critical as Frequency

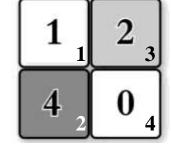
2 possible phases for the higher-frequency components

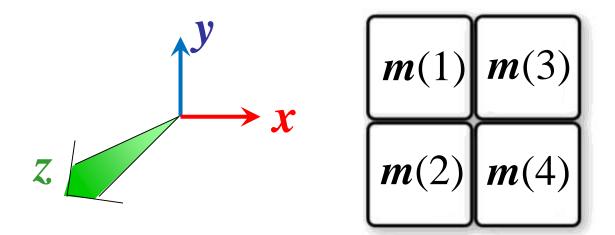


#### PD Map of Thin-Slice 2D, 4-Voxel Patient

after 90° pulse drives M(t) into x-y plane...

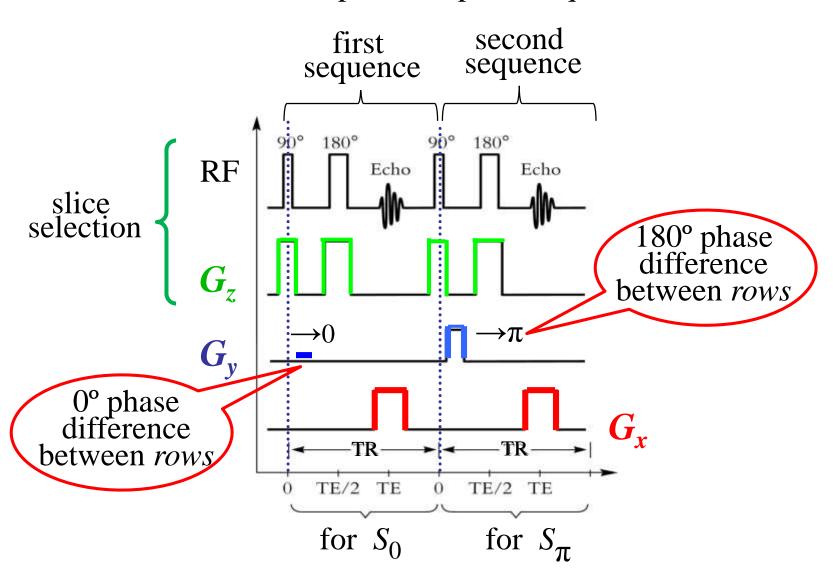
$$M(t) = m(1,t) + m(2,t) + m(3,t) + m(4,t)$$





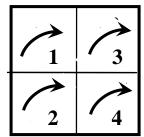
viewed from feet to head

# Spin-Echo, Spin-Warp Sequence for 2×2 Matrix involves 2 spin-echo pulse sequences

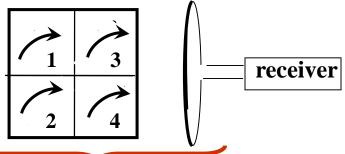


#### No x- or y-Gradients: $G_y = G_x = 0$

after phase encoding









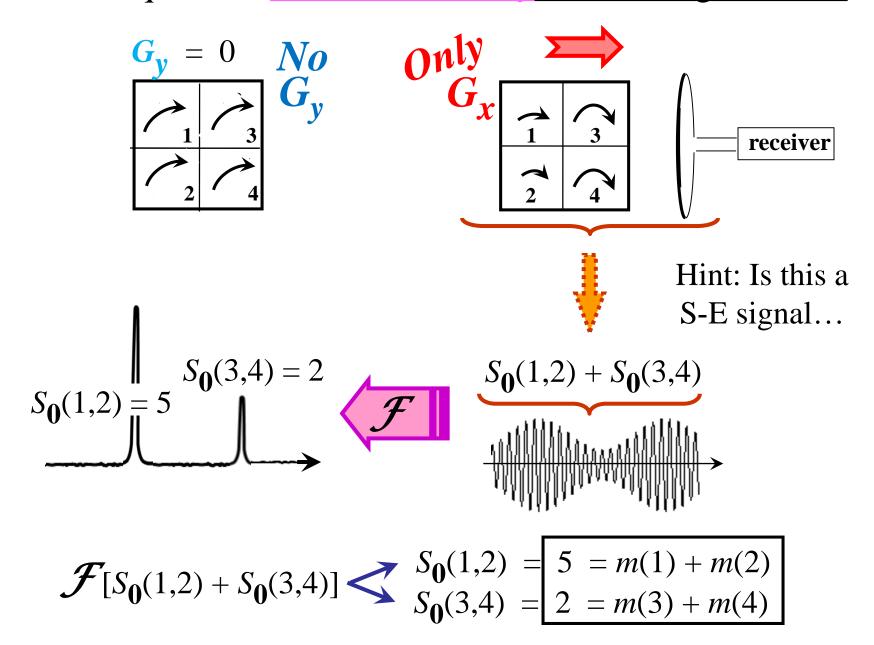
Q: Are we cheating here?

$$S_0(1,2,3,4)$$

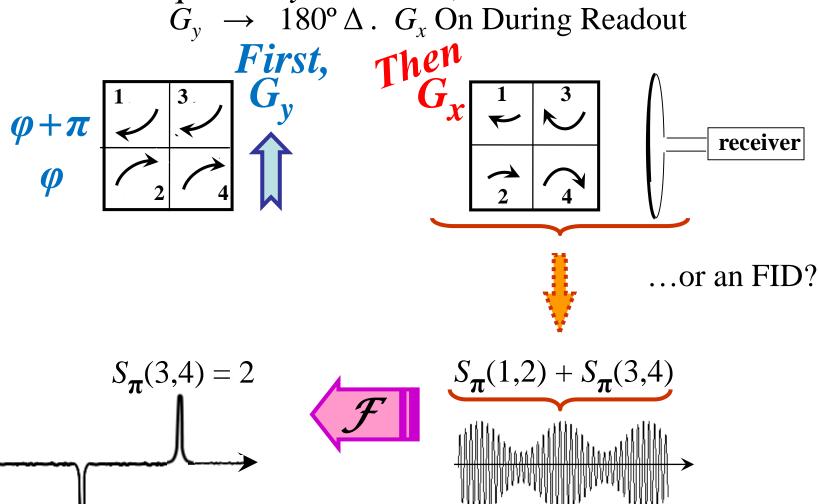
$$S_0(1,2,3,4) = 7 (e.g.)$$
 $f_{\text{Larmor}}$ 

$$\mathcal{F}[S_0(1,2,3,4)] \longrightarrow 7 = m(1) + m(2) + m(3) + m(4)$$

#### 1st S-E Sequence: x-Gradient Only, On During Readout



#### 2<sup>nd</sup> S-E Sequence: y-Gradient, then x-Gradient



$$S_{\pi}(1,2) = -3$$

$$S_{\pi}(1,2) + S_{\pi}(3,4)$$

$$S_{\pi}(1,2) = -3 = m(1) - m(2)$$

$$S_{\pi}(3,4) = 2 = m(3) - m(4)$$

$$S_{\pi}(1,2) = -3 = m(1) - m(2)$$
  
 $S_{\pi}(3,4) = 2 = m(3) - m(4)$ 

#### 4 Equations in 4 Unknowns

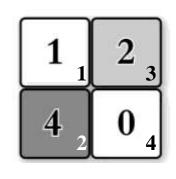
in phase: add
$$S_0(1,2) + S_0(3,4) = \begin{cases} S_0(1,2) = 5 = m(1) + m(2) \\ S_0(3,4) = 2 = m(3) + m(4) \end{cases}$$

$$S_{\pi}(1,2) + S_{\pi}(3,4)$$

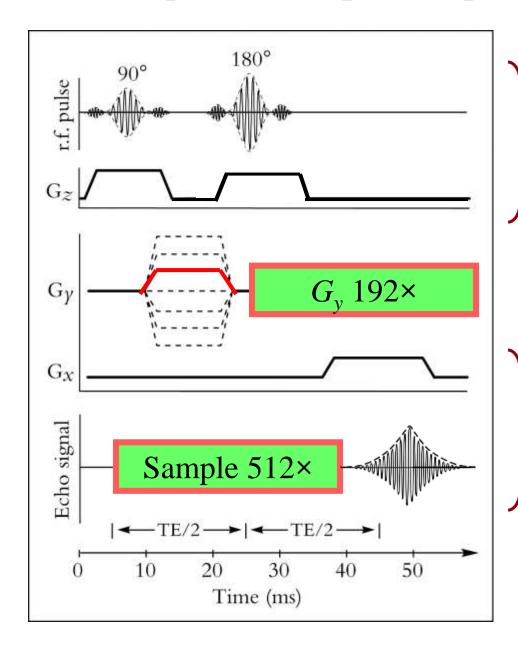
$$S_{\pi}(1,2) = \begin{bmatrix} -3 &= m(1) - m(2) \\ S_{\pi}(3,4) &= 2 &= m(3) - m(4) \end{bmatrix}$$

Solution:  

$$m(1) = 1$$
  
 $m(2) = 4$   
 $m(3) = 2$   
 $m(4) = 0$ 



check: 
$$S_0(1,2,3,4) \xrightarrow{\mathcal{F}} 7 = m(1) + m(2) + m(3) + m(4) !$$



90° pulse with  $G_7$  on selects z-plane and initiates spin-echo

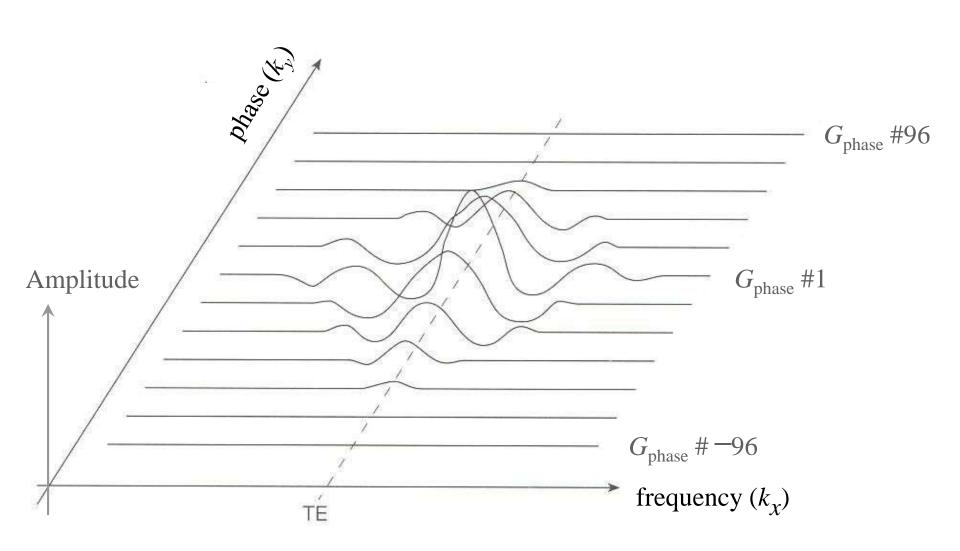
Free precession with  $G_{\mathbf{v}}$  on phase-encodes y-position for current n,  $G_{\mathbf{v}}(n) = n \times G_{\mathbf{v}}(1)$ 

Reading NMR signal with  $G_{\mathbf{r}}$  on <u>frequency-encodes *x*-position</u>

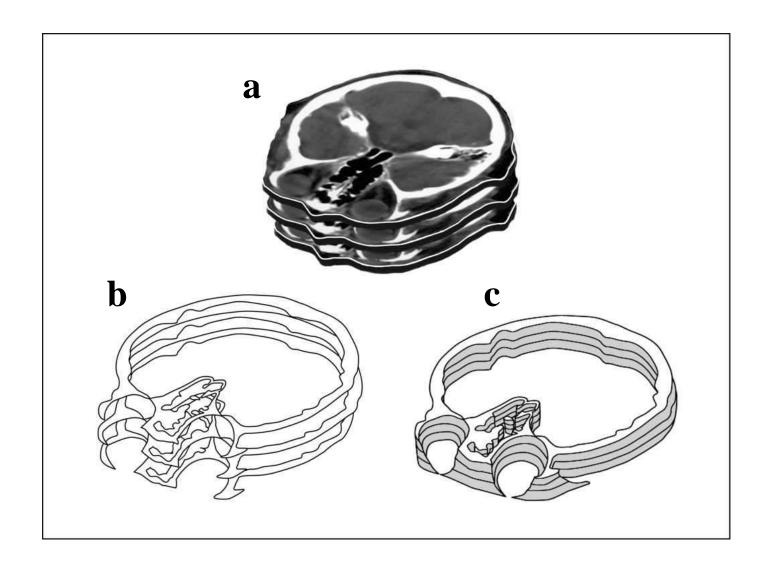
Repeat, but with  $G_{\mathbf{v}}(n+1)$ , n = 1, 2, ..., 192

#### Creation of an MRI 'Image' in 2D k-Space

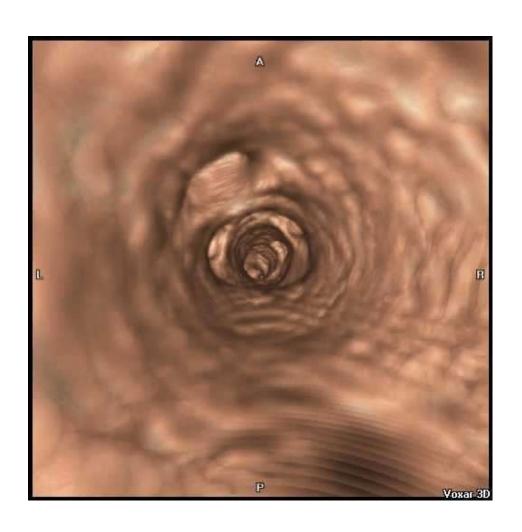
each line in k-space from data obtained during one  $M_{xy}(TE)$  readout; different lines for different phase-encode gradient strength



#### Creating a 3D Image



#### Interior of Bronchus





#### Some Important Topics Not Addressed Here

2D, 3D, 4D Imaging

Inversion Recovery (STIR and FLAIR)

Fast S-E; Gradient Recovery Imaging (GRE, e.g., EPI)

Dynamic Contrast-Agent Enhancement (DCE)

Magnetization Transfer

CNR and Other Quantitative Measures of Image Quality

Parallel-Coil Receive, Transmit; Shim Coils

Magnetic Resonance Angiography (MRA)

**Perfusion Imaging** 

**Diffusion Tensor Imaging (DTI)** 

Functional MRI (f MRI)

Image QA, and ACR Accreditation

MRI/PET, MRI-Elastography, MRI/DSA, MRI/US

Highly Mobile MRI (e.g., for strokes)

**Zero-Quantum Imaging** 

Artificial Intelligence Diagnosis

#### MRI-Guided Radiation Therapy: Tomorrow!

## A Brief Introduction to Magnetic Resonance Imaging

