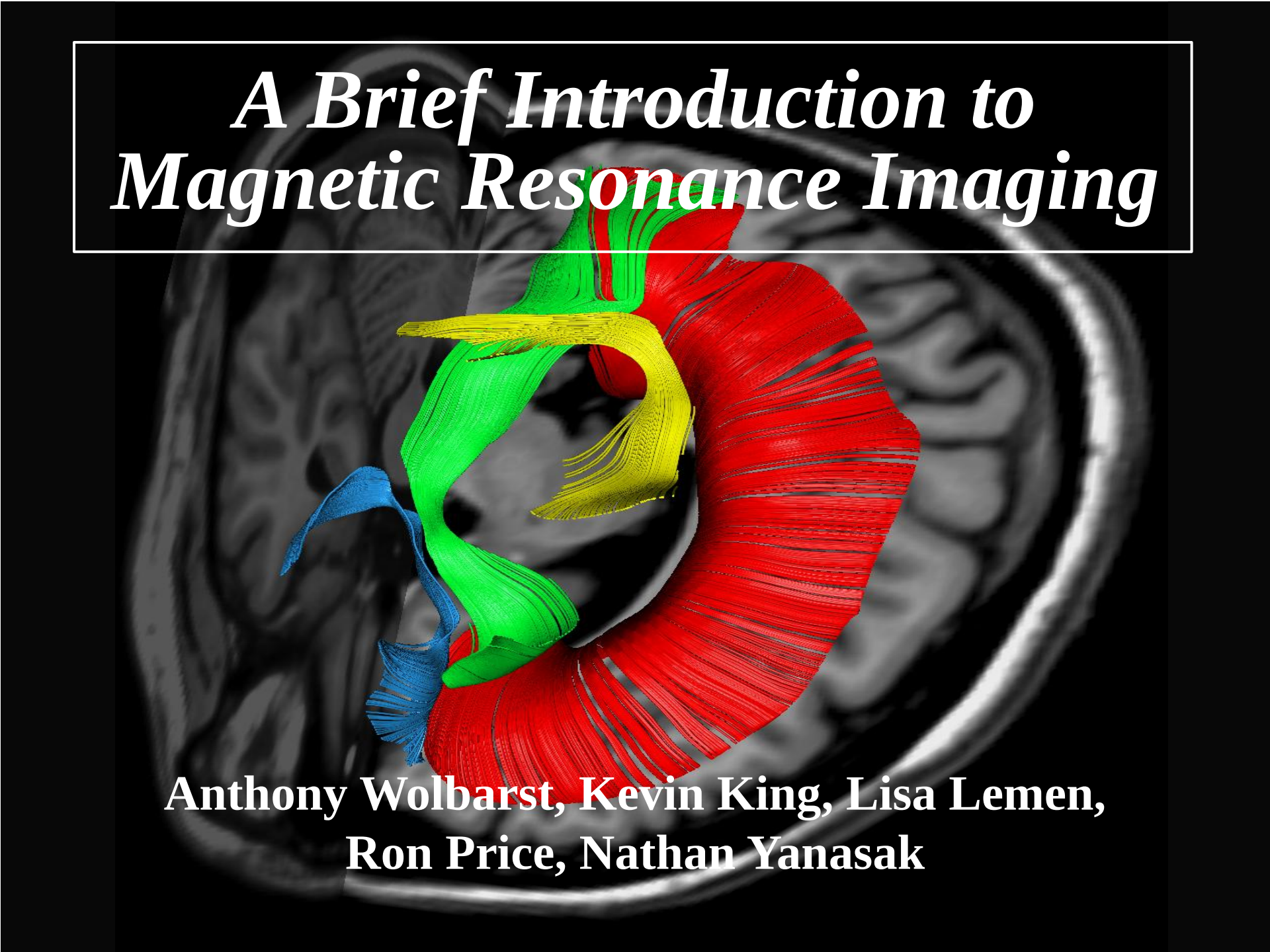
A Brief Introduction to Magnetic Resonance Imaging



Anthony Wolbarst, Kevin King, Lisa Lemen, Ron Price, Nathan Yanasak

Outline for Today

1) Introduction — Wolbarst
   MRI Case Study; ***Caveat!!!***
   "Quasi-Quantum" Nuclear Magnetic Resonance

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2) Net magnetization, $\boldsymbol{m}(x,t)$, of the voxel at x — Lemen
   T1 Spin-Relaxation of $\boldsymbol{m}(x,t)$,
   T1 MRI of the 1D patient
   Sketch of the MRI Device

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3) 'Classical' Approach to NMR — Yanasak
   FID Image Reconstruction, ***k***-Space

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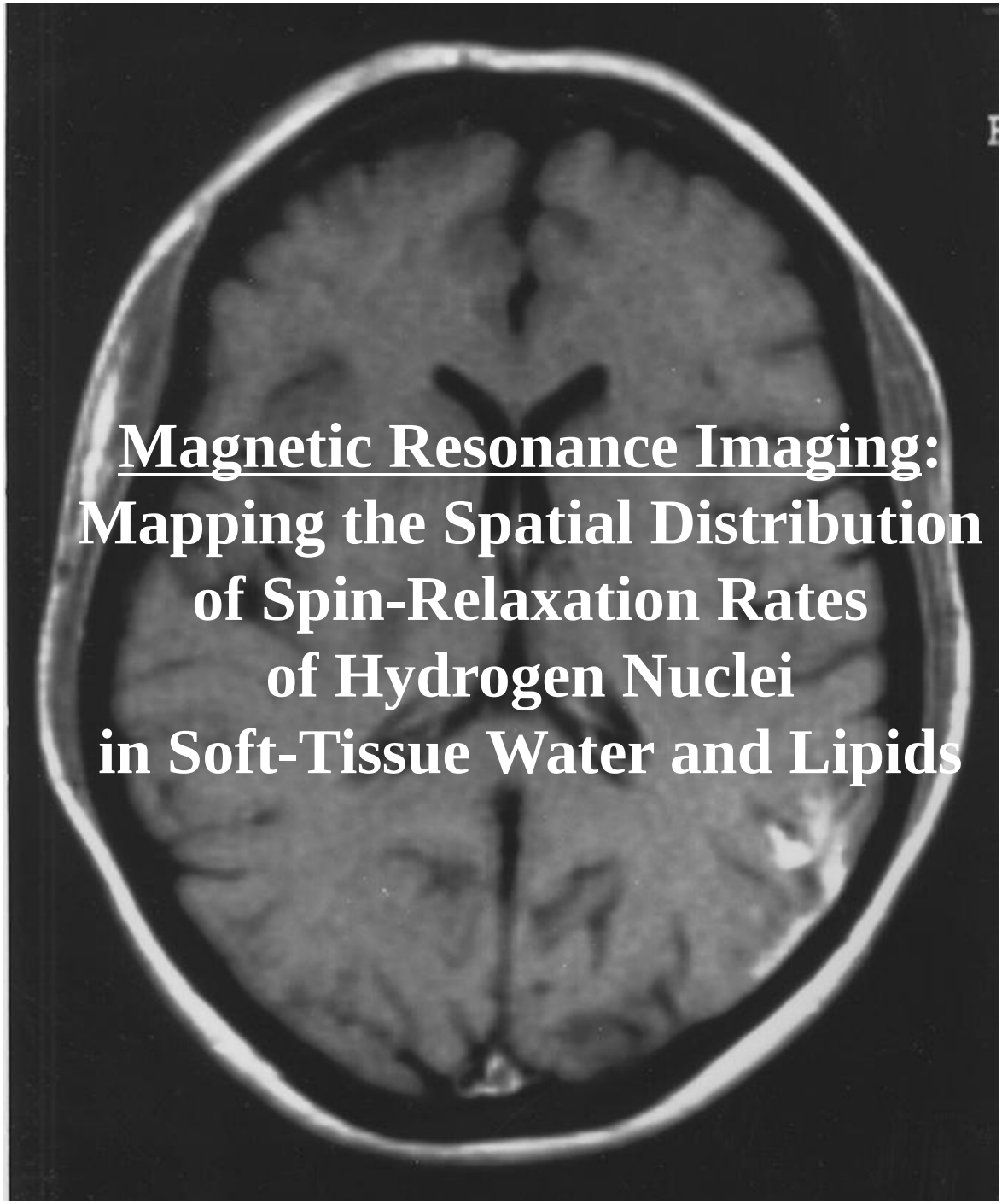
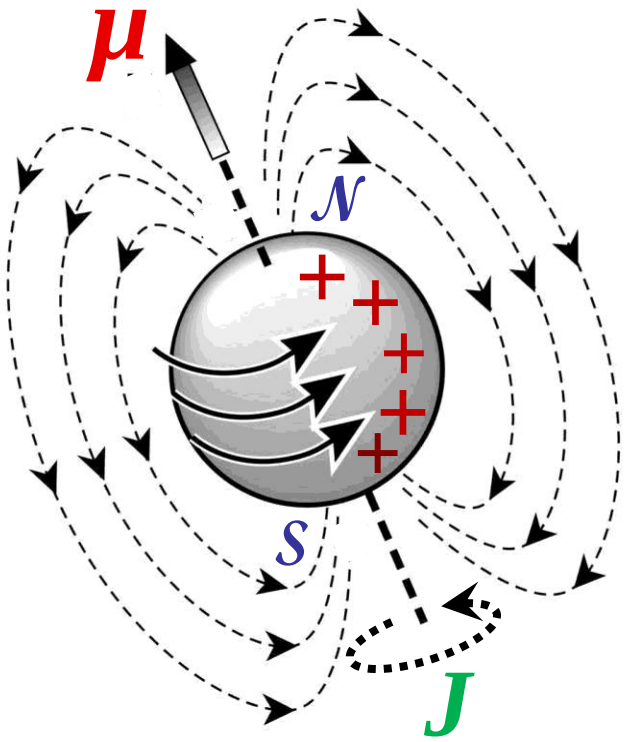
4) Spin-Echo Reconstruction — Price
   T2 Spin-Relaxation
   T1-*w*, T2-*w*, and PD-w S-MRI
   Spin-Echo / Spin-Warp in 2D
   Some Important Topics Not Addressed Here


Magnetic Resonance Imaging:
Mapping the Spatial Distribution
of Spin-Relaxation Rates
of Hydrogen Nuclei
in Soft-Tissue Water and Lipids

Part 1: "Quasi Quantum" NMR and MRI



Introduction

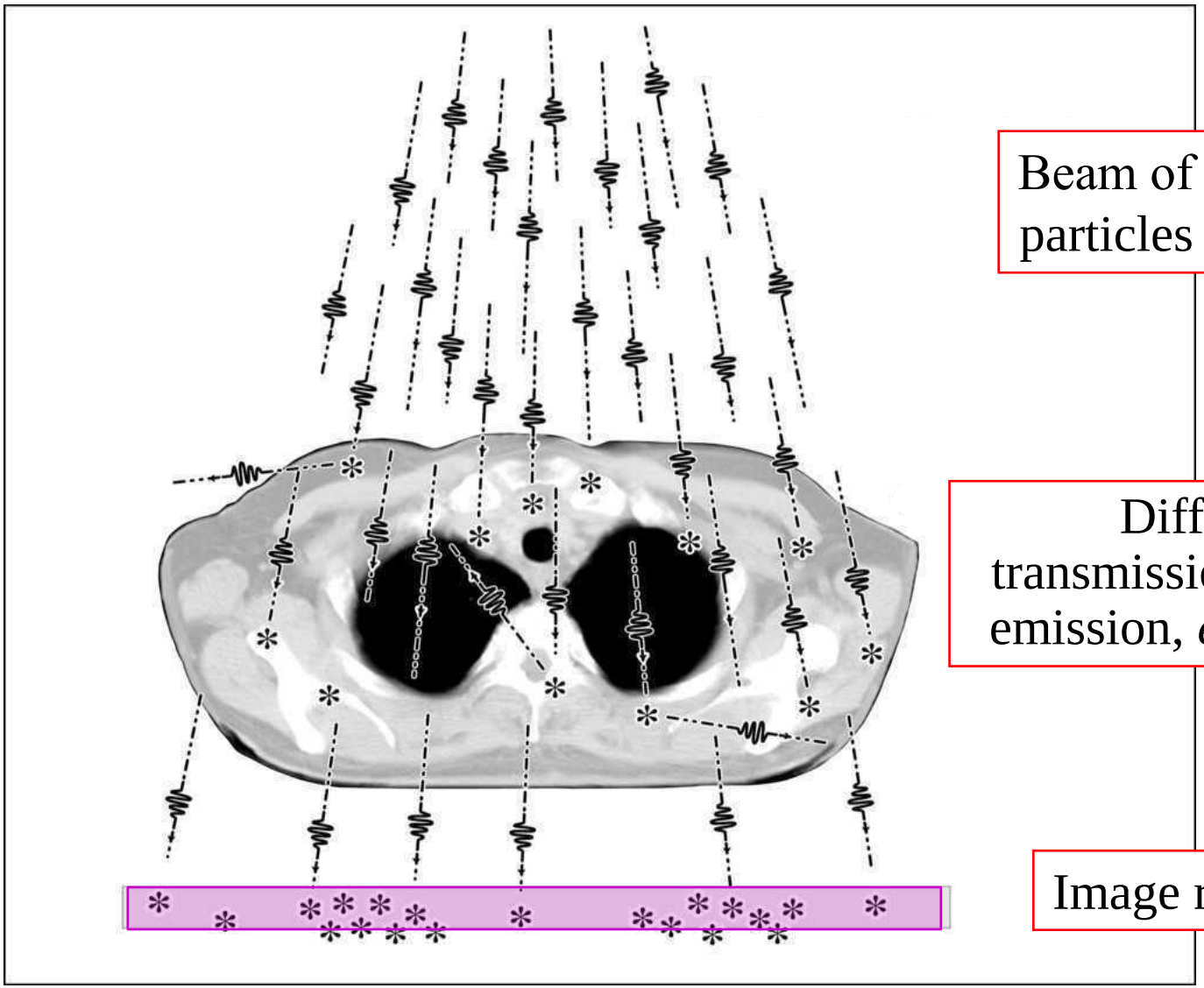
MRI Case Study; ***Caveat!!!***

"Quasi-Quantum" NMR and MRI

Introduction

Medical Imaging:

differential interactions of probes with different tissues



Beam of ‘probes’: particles or waves

Differential transmission, reflection, emission, *etc*., of probes

Image receptor

Forms of Contrast Between Good and Bad Apples

produced and *detected* thru different bio/chemico/physical processes

The Future: Contrast Mechanisms!

Color

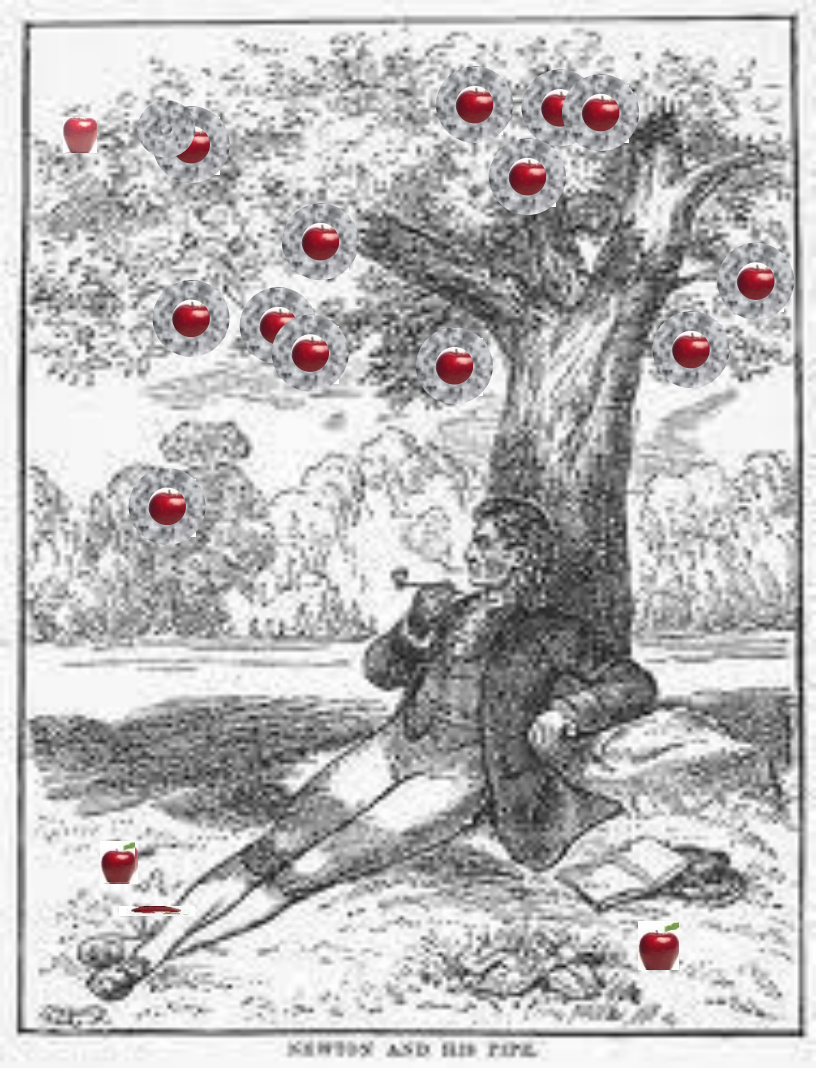
Smell

Taste

Texture

Holes

etc.

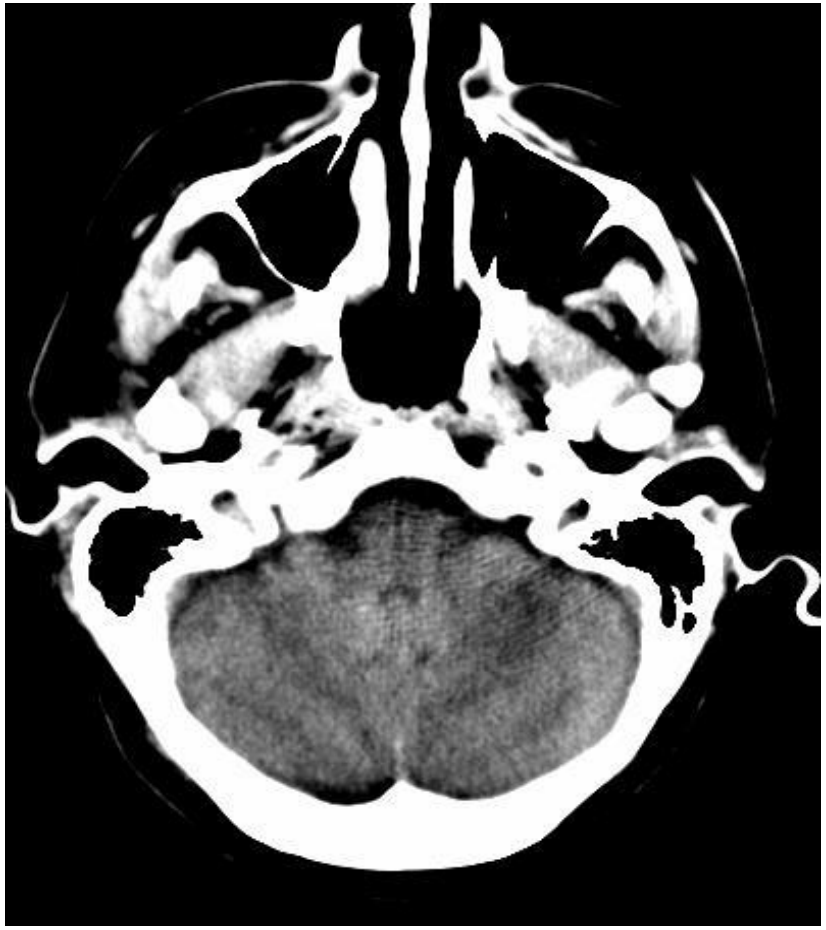


How Imaging Modalities Detect Tissue Contrast…

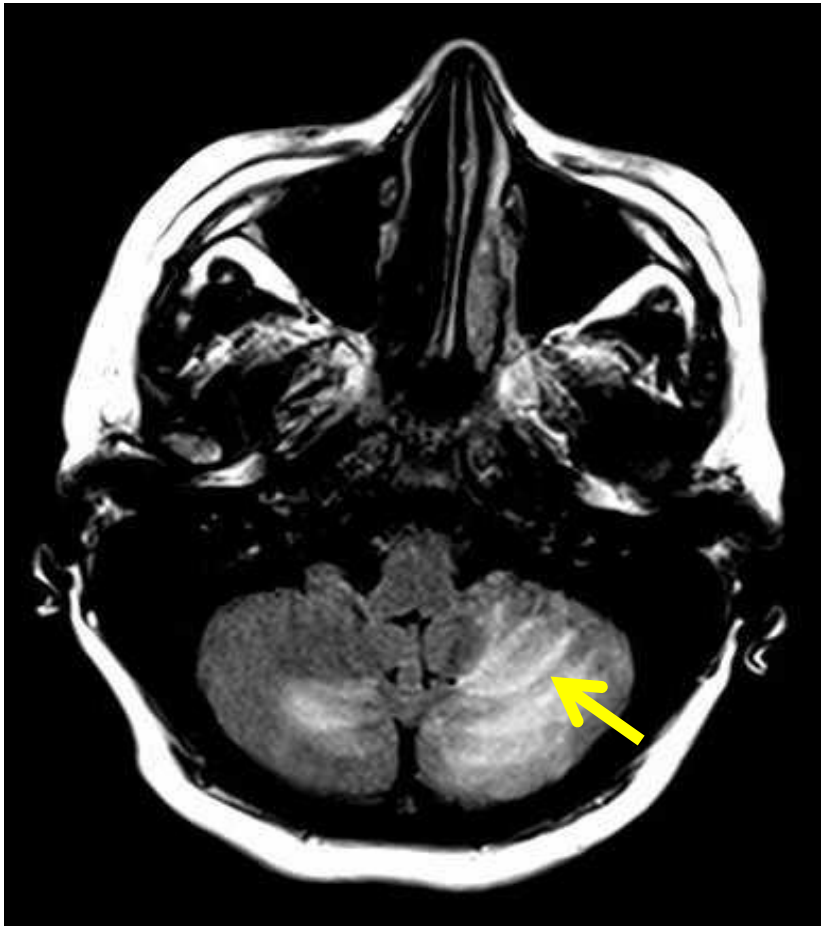
Modality	Probe / Signal	Detector	Source of *Contrast*: differences in…
Planar R/F X-ray CT	X-rays passing through body	II+CCD, AMFPI; GdO, *etc.*, array	$\int_S \boldsymbol{\mu}(\rho, Z, kVp)\, ds$
Nuc Med, SPECT; PET	Gamma-rays from body; 511 keV	NaI single crystal; multiple NaI; LSO array	Radiopharmaceutical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

CT *vs.* MRI Soft Tissue Contrast

CT



MRI



Posterior reversible encephalopathy syndrome (PRES), edematous changes

Magnetic Resonance Imaging

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

Selection of contrast type with pulse sequences

Most reflect rotations, flow, *etc*., of water/lipid molecules; yield distinct, unique maps of anatomy, physiology

No ionizing radiation

Risks from intense magnetic fields, RF power

Expensive

Technology complex, challenging to learn;

Much of biophysics little understood. But….

MRI Case Study; and ***Caveat!!!***

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

Case Study

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

Physical examination 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.

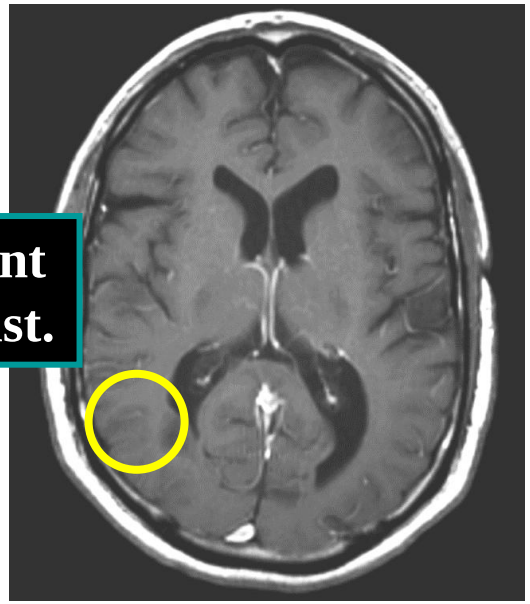
CT indicates a lesion in the *right posterior temporo-occipital* region, adjacent to *occipital horn of right lateral ventricle*. MRI for more information.

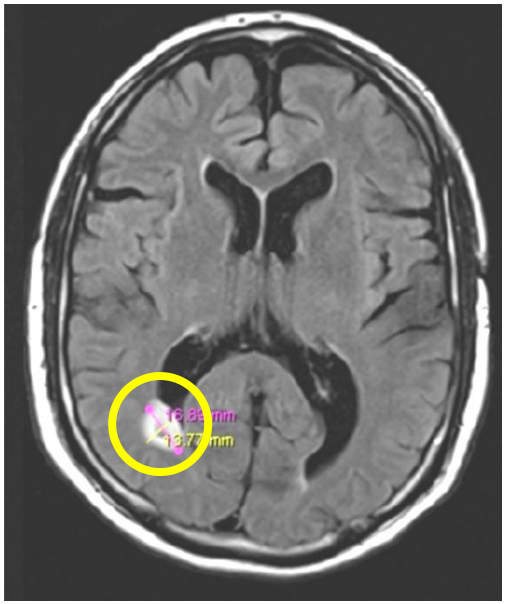
Principal concern: *Vision for reading*.

Lesion: Right Posterior Temporo-Occipital Region,

adjacent to occipital horn of right lateral ventricle

T1-*w*

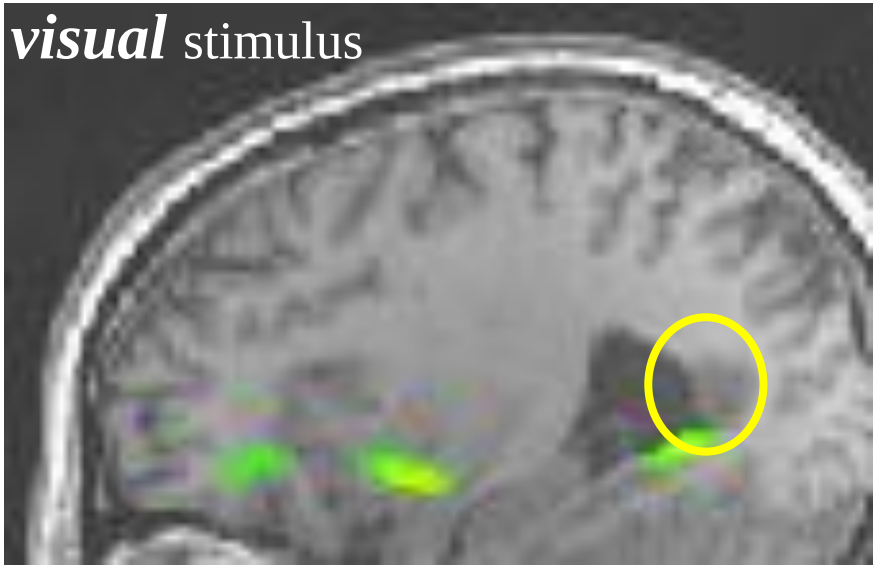




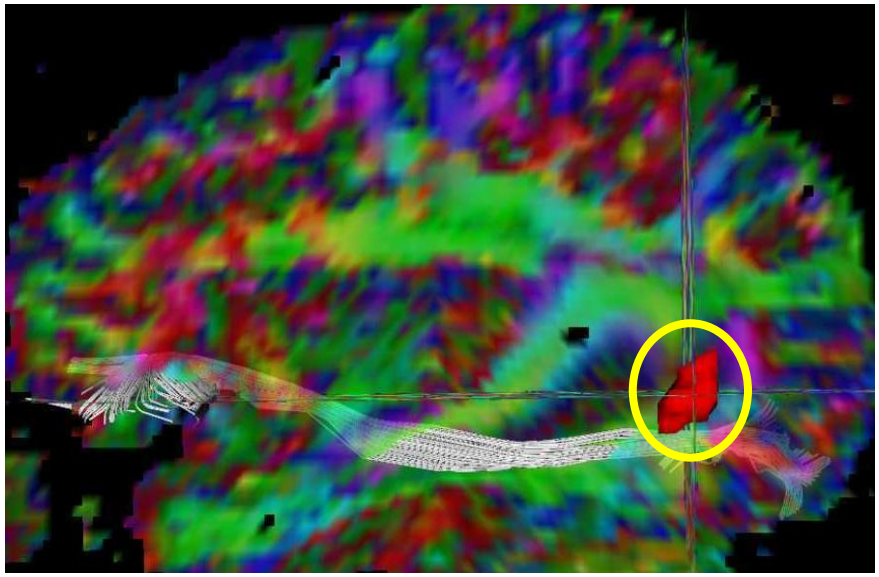
FLAIR

No enhancement with Gd contrast.

f MRI

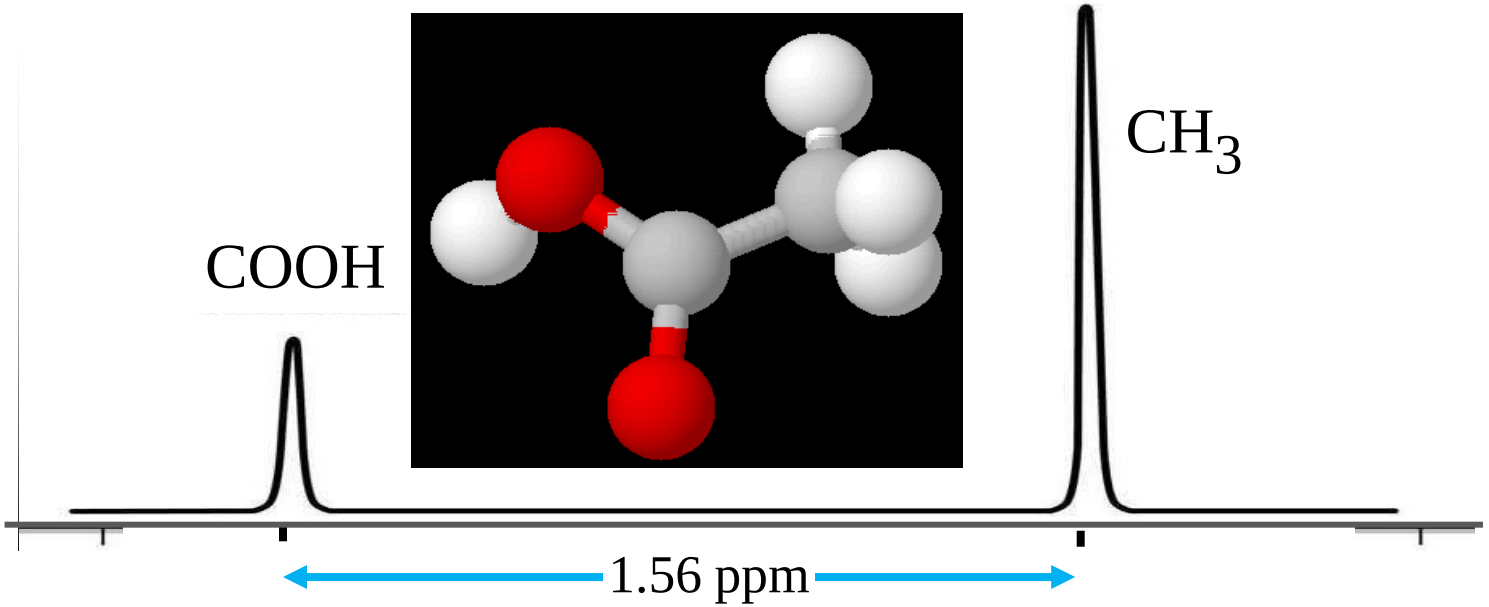


DTI

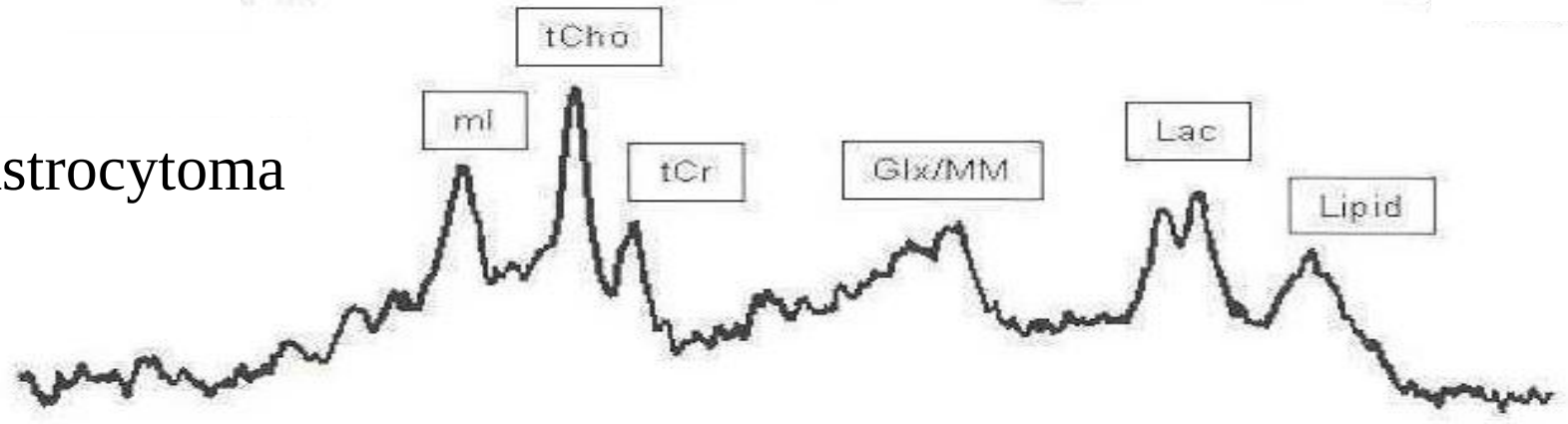


Chemical Shift and Non-invasive MRS

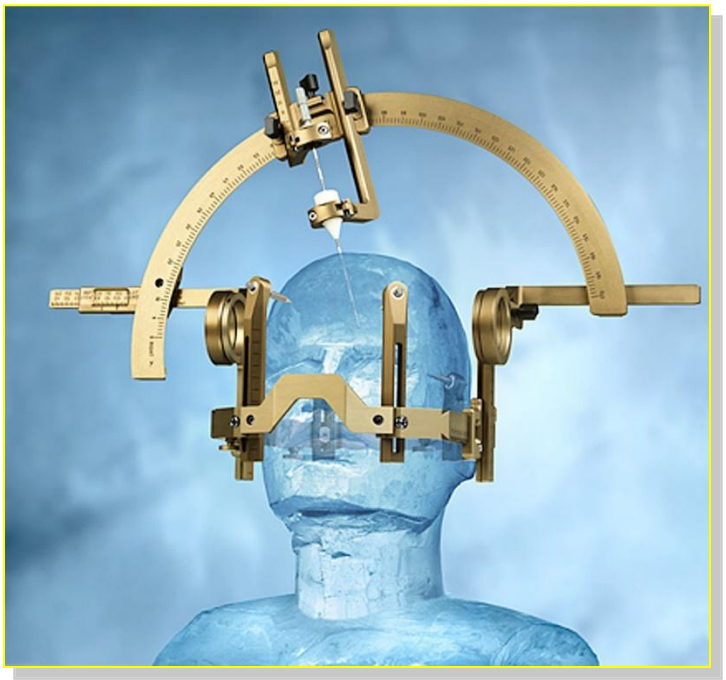
acetic acid



Astrocytoma



MRI-Guided Needle Biopsy



Caveat!!!

Extreme Danger

How *Not* to Clean a Magnet

strong gradient fields outside bore

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 – First and Foremost, Restrict Entry!

within / with patient, others	*in / into imaging suite*
aneurysm clip, shrapnel	hemostat, scalpel, syringe
cochlear implant, prostheses	O_2 bottle, IV pole
artificial heart valve	scissors, pen, phone, laptop
stent, permanent denture	tool, tool chest
defibrillator, pacemaker,	wheelchair, gurney
electrodes, nerve stimulator	ax, fire extinguisher
medical infusion pump	gun, handcuffs
drug-delivery patch, tattoo	cleaning bucket, mop

“Quasi-Quantum” NMR and MRI

Swept-frequency NMR in a tissue voxel

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

Magnetization, $\boldsymbol{m}(x,t)$, in the voxel at $\boldsymbol{x}$

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

quantum spin-state function for hydrogen nucleus

$|\psi\rangle$

start with this

Simple QM

$|\uparrow\rangle, |\downarrow\rangle$

transitions between
spin-up-, spin-down states

f_{Larmor} , m_0 , T1

oversimplified;
like Bohr atom

Classical Bloch Eqs.

expectation values in voxel

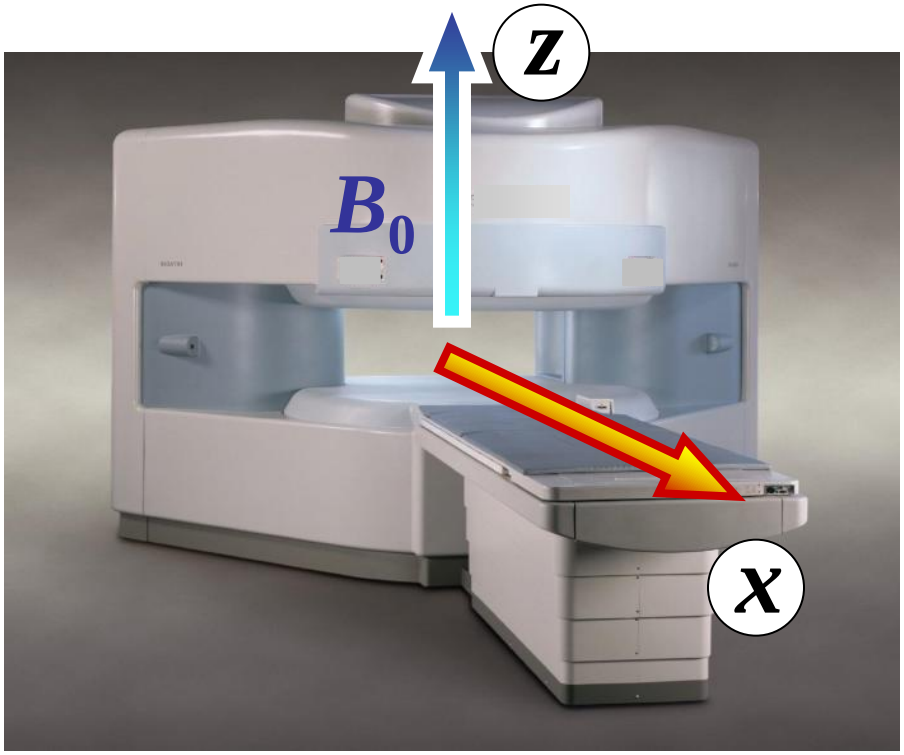
precession, nutation of
net voxel magnetization, $\boldsymbol{m}(t)$

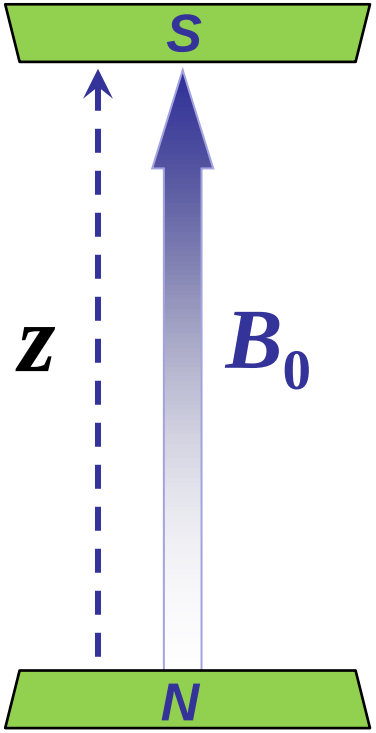
f_{Larmor} , T2 , 2D, $\boldsymbol{k}$-space

exact, for expectation values;
from full QM

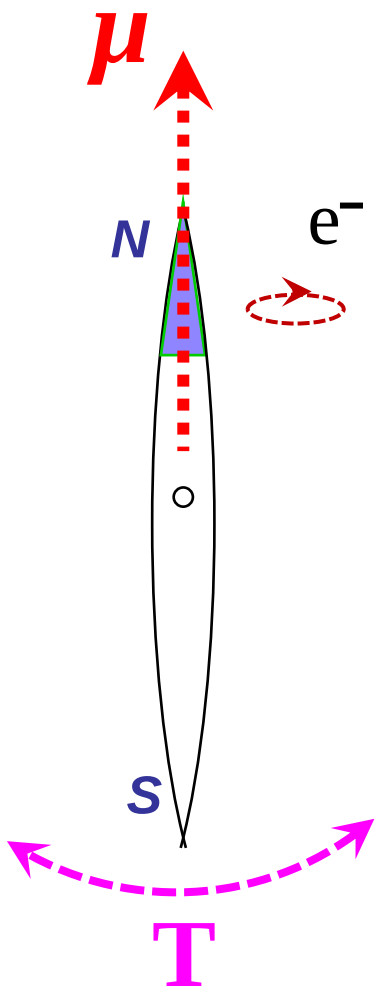
‘Open’ MRI Magnet

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

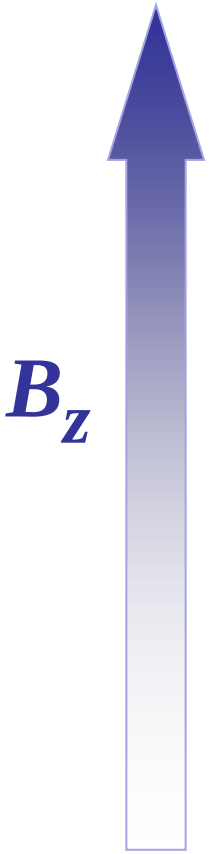


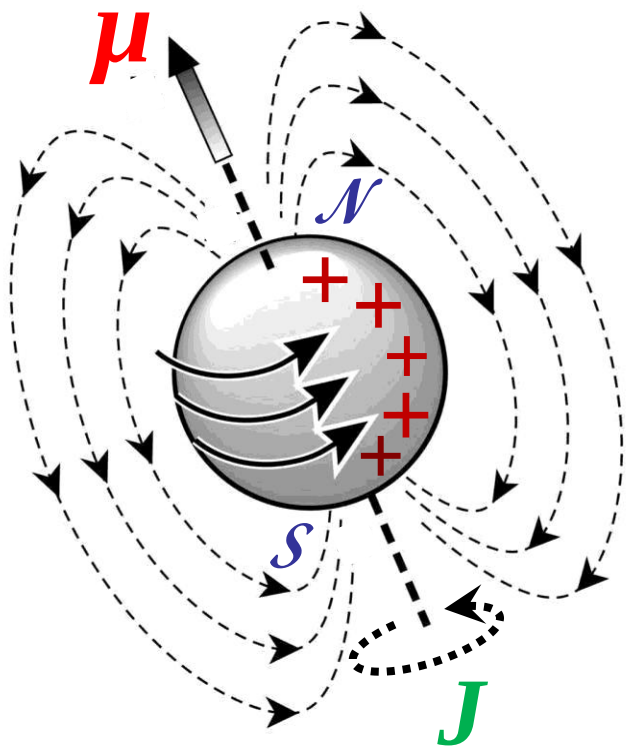


Moving Charge Produces Magnetic Field (*e.g.*, mag. dipole); Magnetic Dipole Tends to Align in External Field



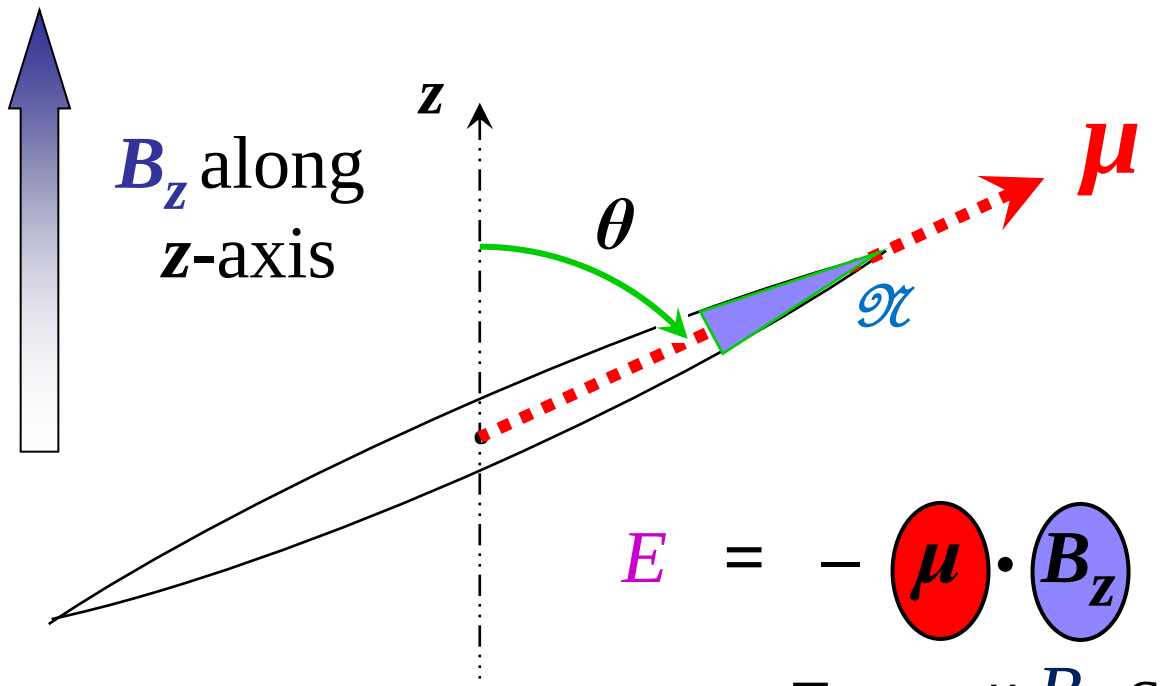
mechanical *relaxation*





simplified QM picture
proton

Energy to Flip Over Needle with Magnetic Moment μ in B_z



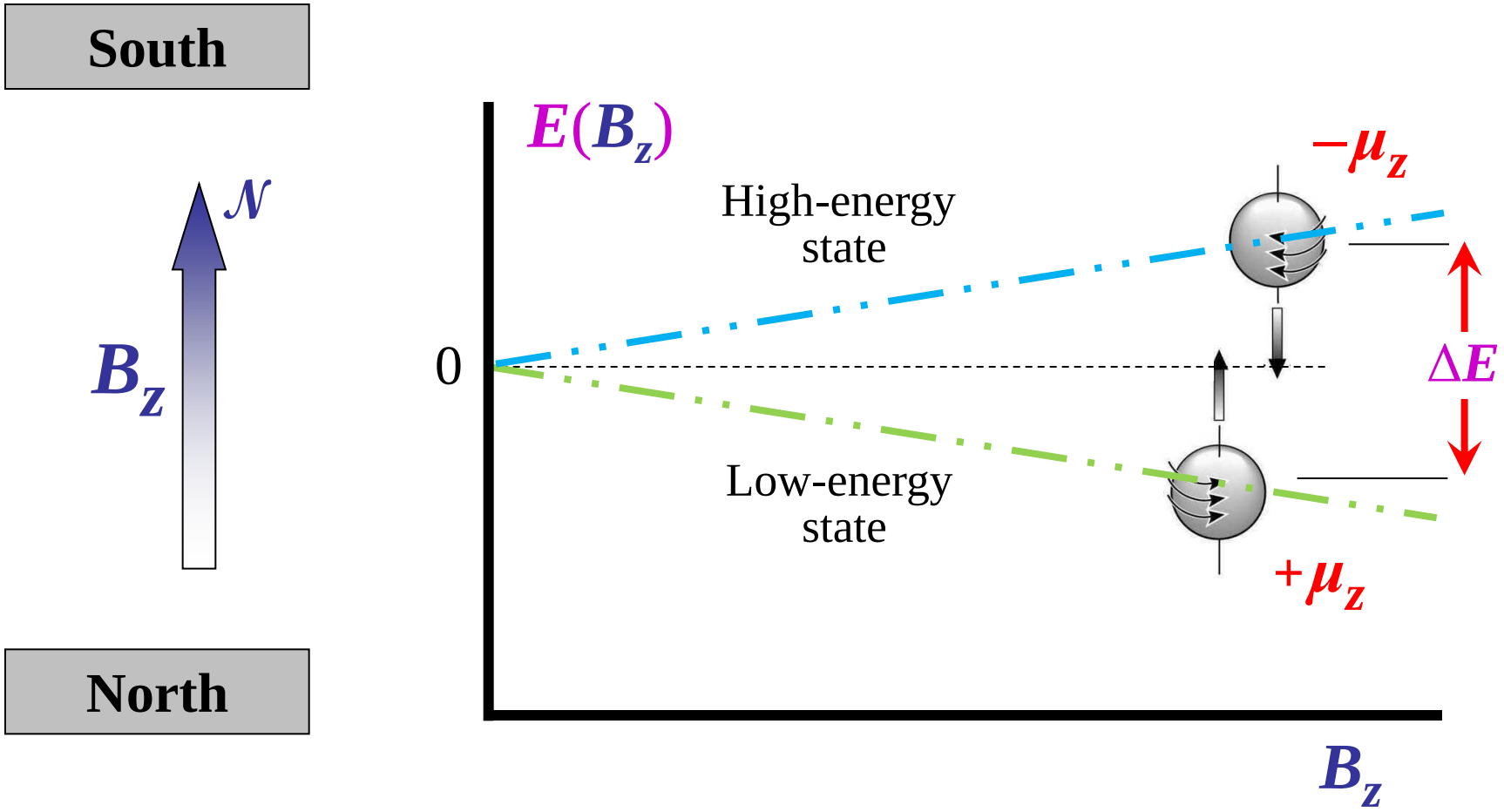
$$E = -\mu \cdot B_z$$

$$= -\mu B_z \cos\theta$$

$$\Delta E_{180^\circ} = \pm 2 \mu B_z$$

Nuclear Zeeman Splitting for Proton: $\Delta E = \pm 2 \mu_z B_z$

$\boldsymbol{\mu}$ points only along or against z



Swept-frequency NMR in a tissue voxel

Nuclear Magnetic Resonance

proton $\Delta E_{180°} = 2\,\mu_z B_z$

photon $E = hf$

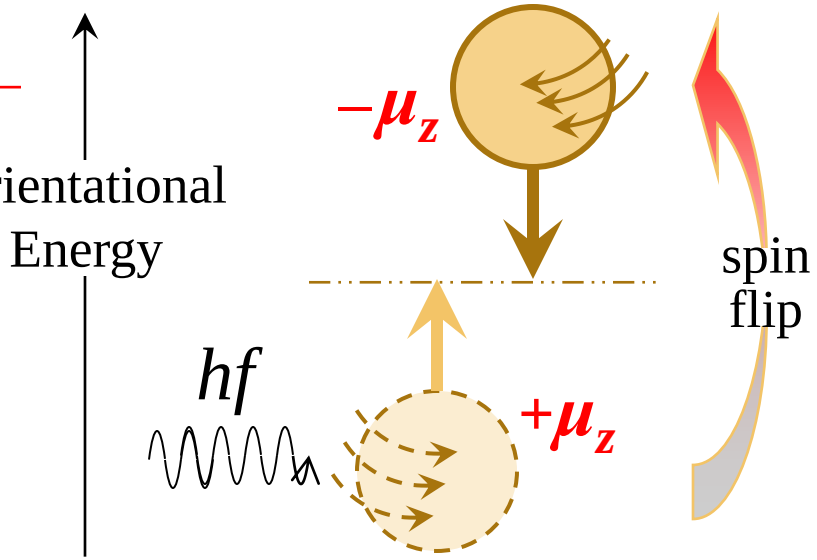
$$hf_{\text{Larmor}}(B_z) = 2\mu_z B_z$$

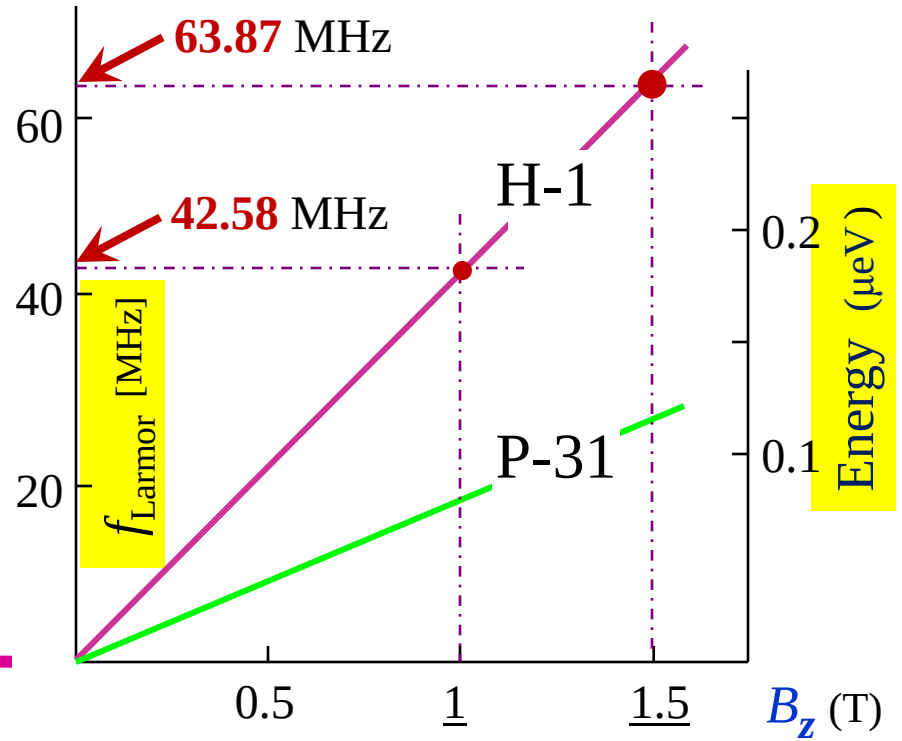
$$2\pi f_{\text{Larmor}} \equiv \omega = \gamma B_z$$

for ***water*** protons,

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

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





SAMS Question

I-1. For protons, $f_{\text{Larmor}} = 42.58$ MHz at 1 T. What is it at 1.5 T?

8%	a. 42.58 MHz
2%	b. 28.36 MHz
84%	c. 63.87 MHz
3%	d. 21.39 MHz
2%	e. Cannot be determined from this info.

SAMs Q:

I-1. For protons, f_{Larmor} = 42.58 MHz at 1 T. What is it at 1.5 T?

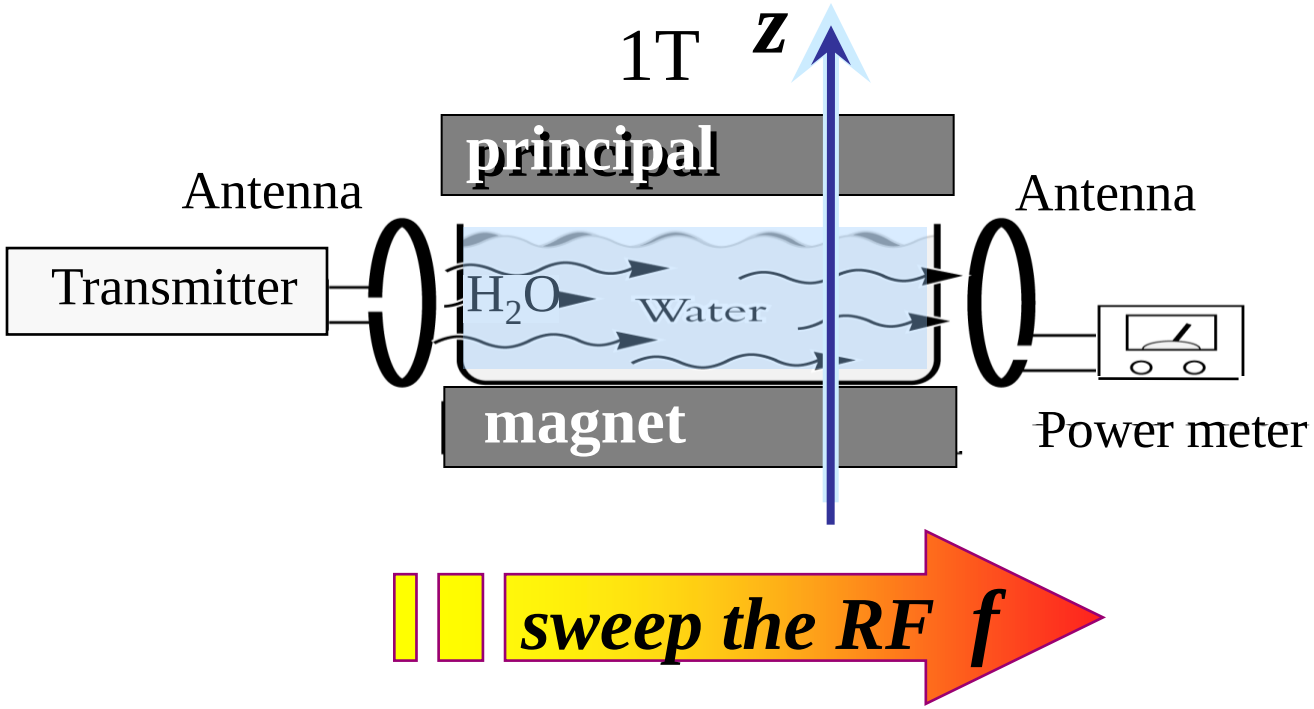
(a) 42.58 MHz
(b) 28.36 MHz
(c) 63.87 MHz
(d) 21.39 MHz
(e) Cannot be determined from this info.

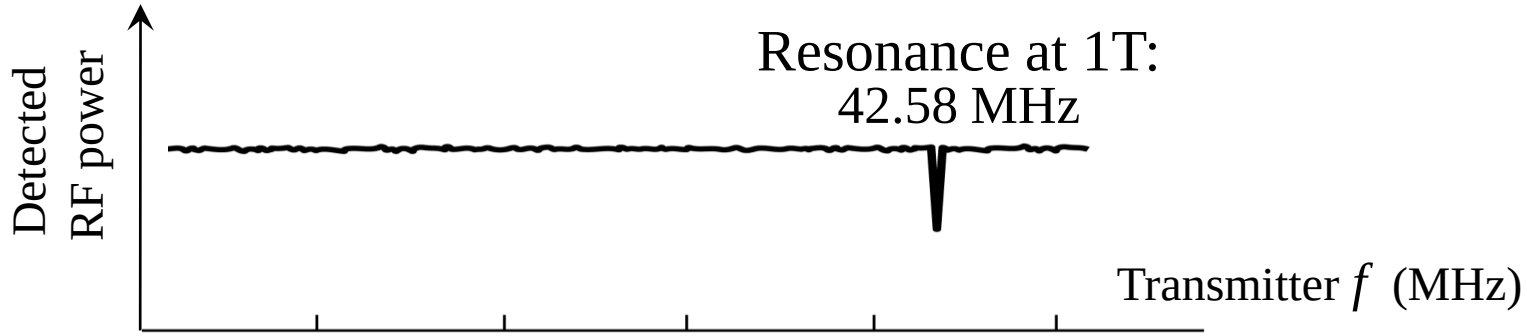
Answer: (c). $f_{\text{Larmor}} = 42.58\, B_z = 42.58$ MHz/T × 1.5 T
= 63.87 MHz

Ref: "Medical Imaging: *Essentials for Physicians*", A.B. Wolbarst, P. Capasso, and A. Wyant. Wiley-Blackwell (2013), p. 314. (MPP booth)

NMR Gedanken-Experiment on Water, 1T

monochromatic RF power absorption at (and only at) $f_{\text{Larmor, } H_2O}$





SAMS Question

I-2. If the field homogeneity of a 1.5 T scanner is measured to be one part per million (1 ppm), what will be the approximate spread in resonant frequencies?

16% a. 1.5 Hz

6% b. 42.58 MHz

8% c. 42.58 kHz

11% d. 63.87 MHz

60% e. 63.87 Hz

SAMs Q:

I-2. If the field homogeneity of a 1.5 T scanner is measured to be one part per million (ppm), what will be the approximate spread in resonant frequencies?

(a) 1.5 Hz
(b) 42.58 MHz
(c) 42.58 kHz
(d) 63.87 MHz
(e) 63.87 Hz

Answer: (e). $\Delta f_{\text{Larmor}} = (42.58\times10^{6}\ \text{MHz/T}) \times (1.5\ \text{T}) \times (10^{-6})$
$= 63.87$ Hz

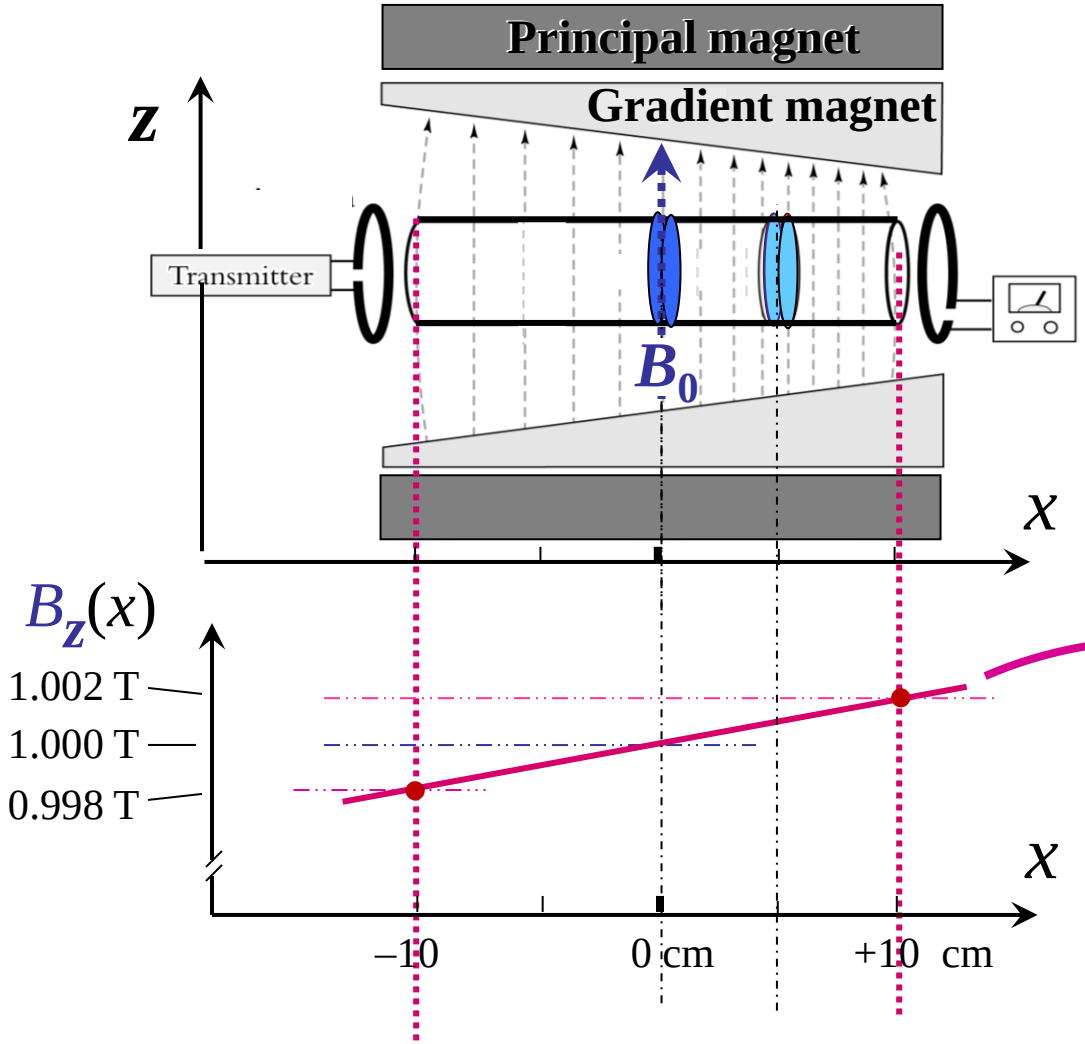
Ref: "Medical Imaging", A.B. Wolbarst *et al.*, Wiley-Blackwell (2013), p. 319.

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

1D Phantom in Principal B_0 and Gradient G_x

$$B_z(x) - B_0 = G_x \times x$$

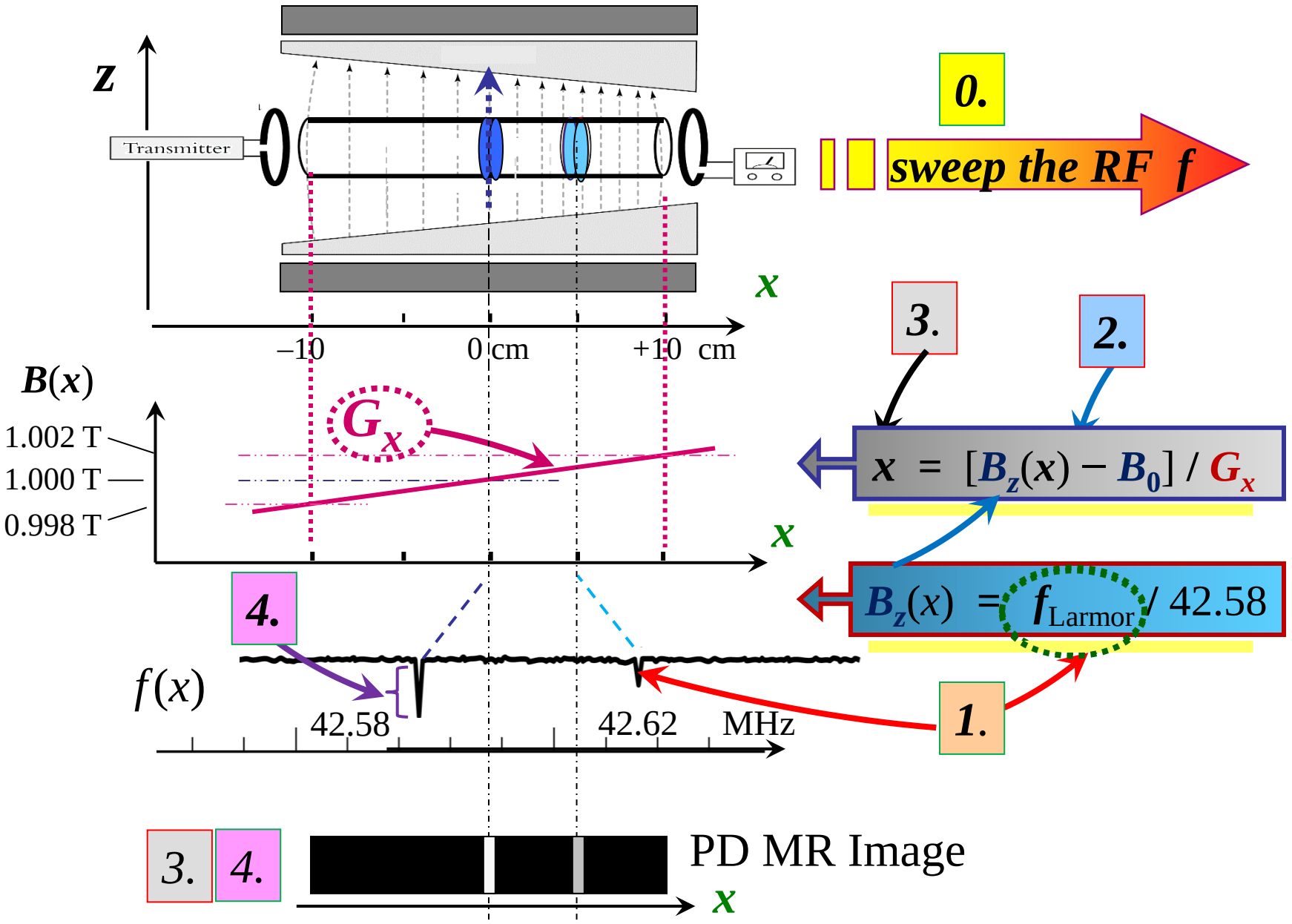
$$G_x \equiv dB_z(x)/dx$$



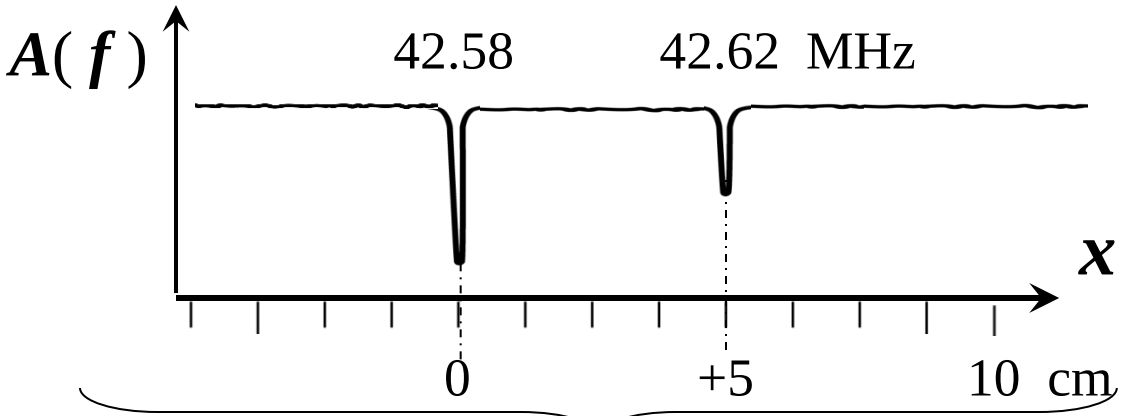
$$x = [B_z(x) - B_0] / G_x$$

$(1.002\text{T} - 0.998\text{T}) / 0.20\text{m}$
$= 20\ \text{mT/m}$

1D PD-MRI: $f_{\text{Larmor}} = f_{\text{Larmor}}[B_z(x)]$



Summary: PD MRI on 1D Patient



1. f_{Larmor} of NMR Peak

4. Amplitude of NMR Peak, $A(f)$

2. $B_z(x)$ — PD(x)

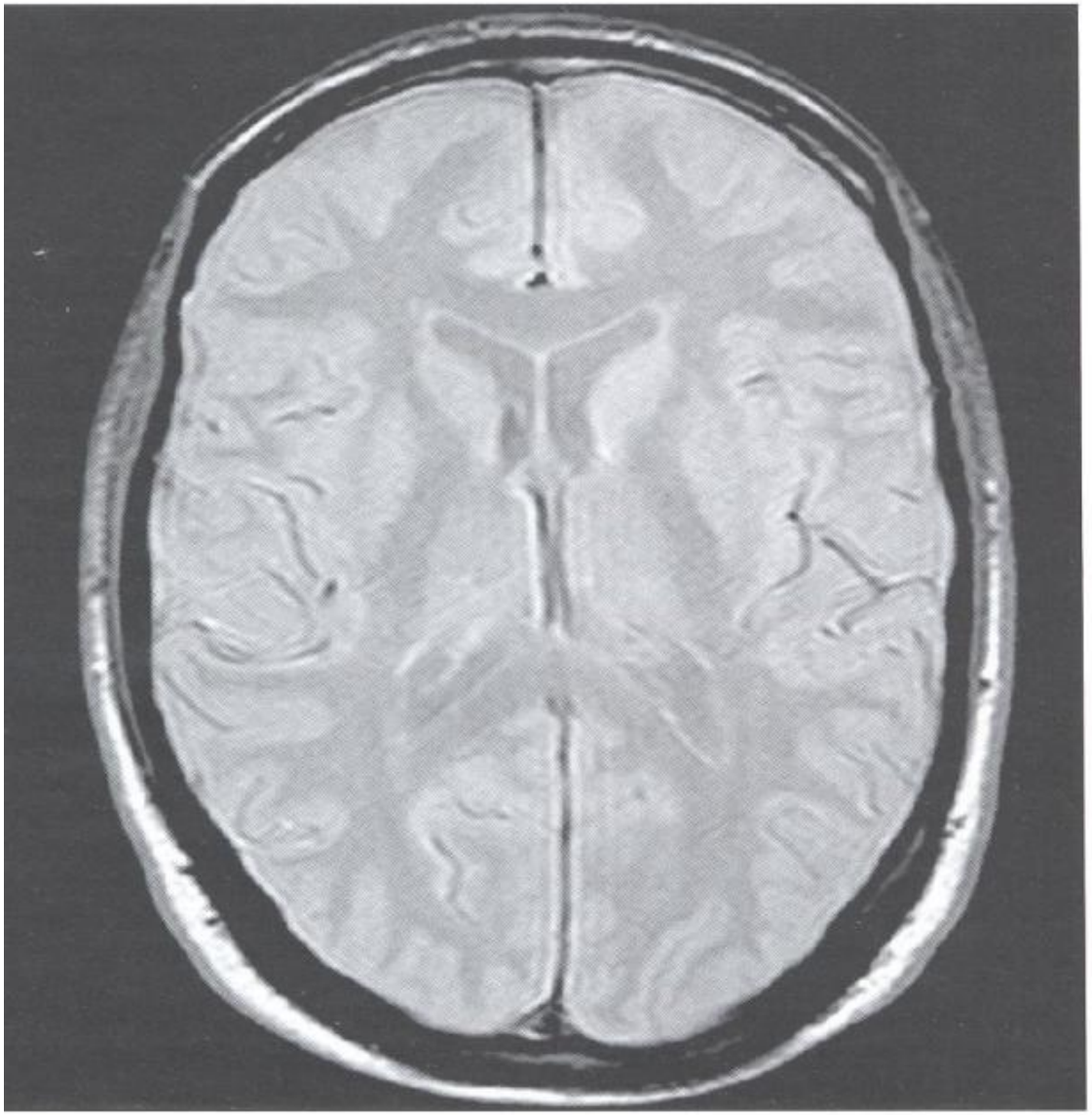
3. Voxel Position, x — Pixel Brightness, PD

4′.

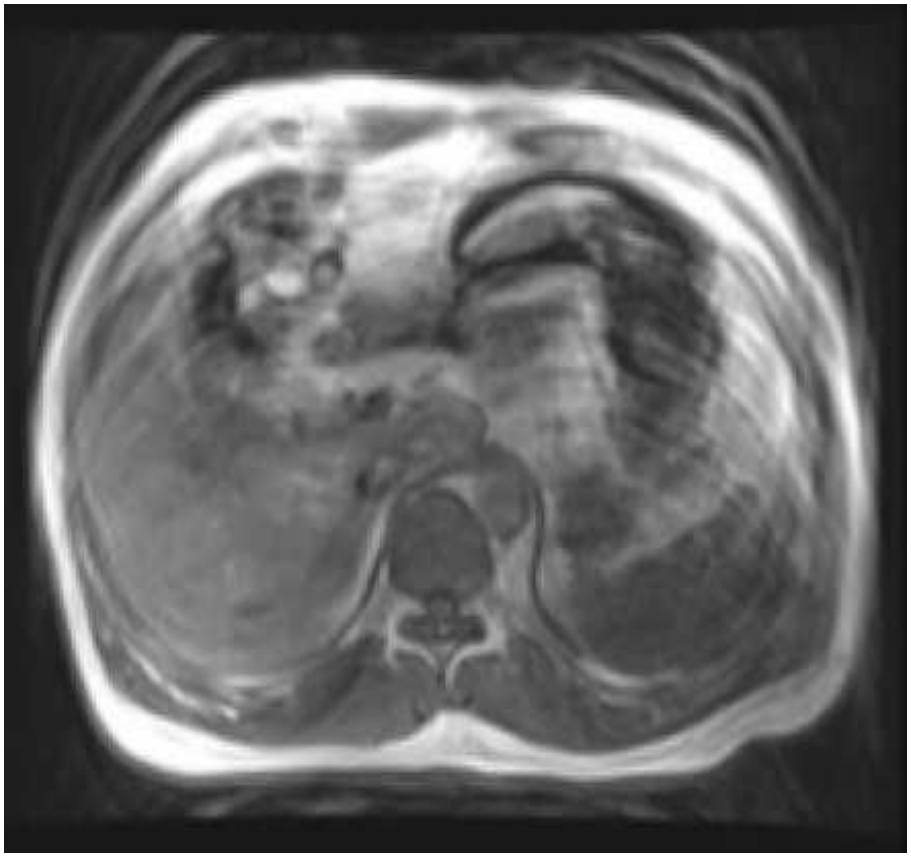
MRI

Proton-Density MRI

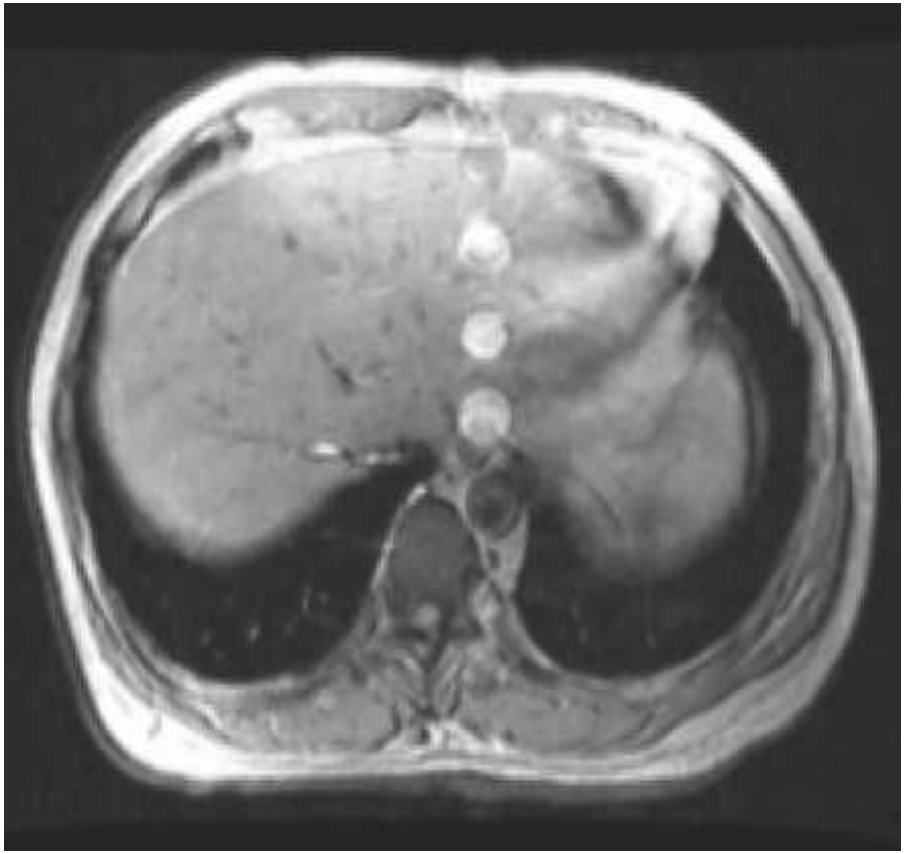
contrast from differences in PD



Two MRI Motion Artifacts



respiration



aortic pulsation

SAMS Question

I-3. The net magnetic field $B_z(x)$ is measured to be 0.999 T at $x = -0.05$ m and 1.001 T at +5 cm. What is G_x?

15% a. 0.01 T/m

29% b. 0.02 T/m

38% c. 20 mT/m

13% d. 0.2 T/cm

5% e. 0.2 T/m

SAMs Q:

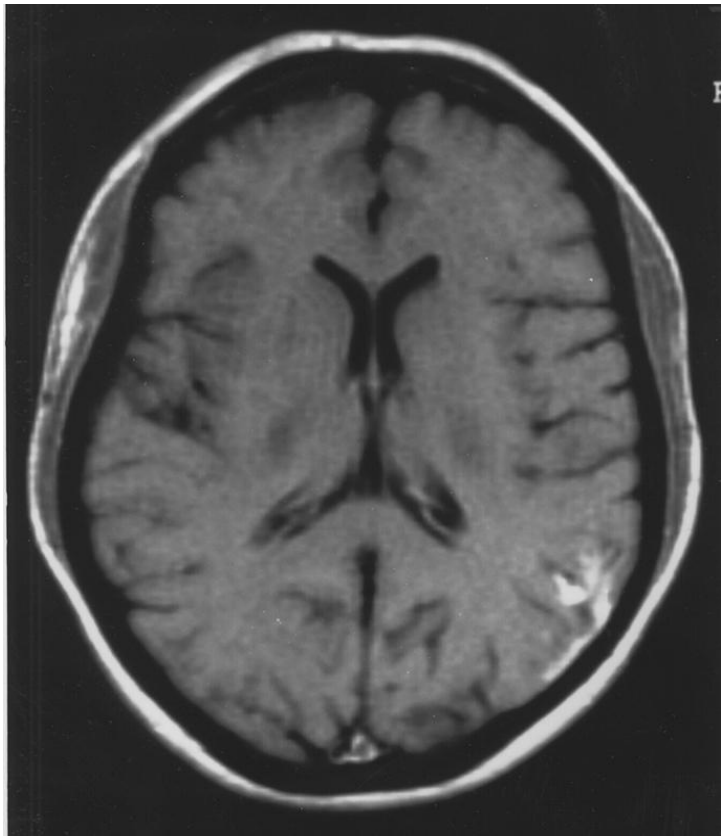
I-3. The net magnetic field $B_z(x)$ is measured to be 0.999 T at $x = -0.05$ m and 1.001 T at +5 cm. What is G_x?

(a) 0.01 T/m
(b) 0.02 T/m
(c) 20 mT/m
(d) 0.2 T/cm
(e) 0.2 T/m

Answer: (c). 20 mT/m. $\Delta B_z(x) / \Delta x = 2 \text{ mT} / 0.1 \text{ m}$

Ref: "Medical Imaging", A.B. Wolbarst *et al.*, Wiley-Blackwell (2013), p. 318.

Part 2: Magnetization & Relaxation

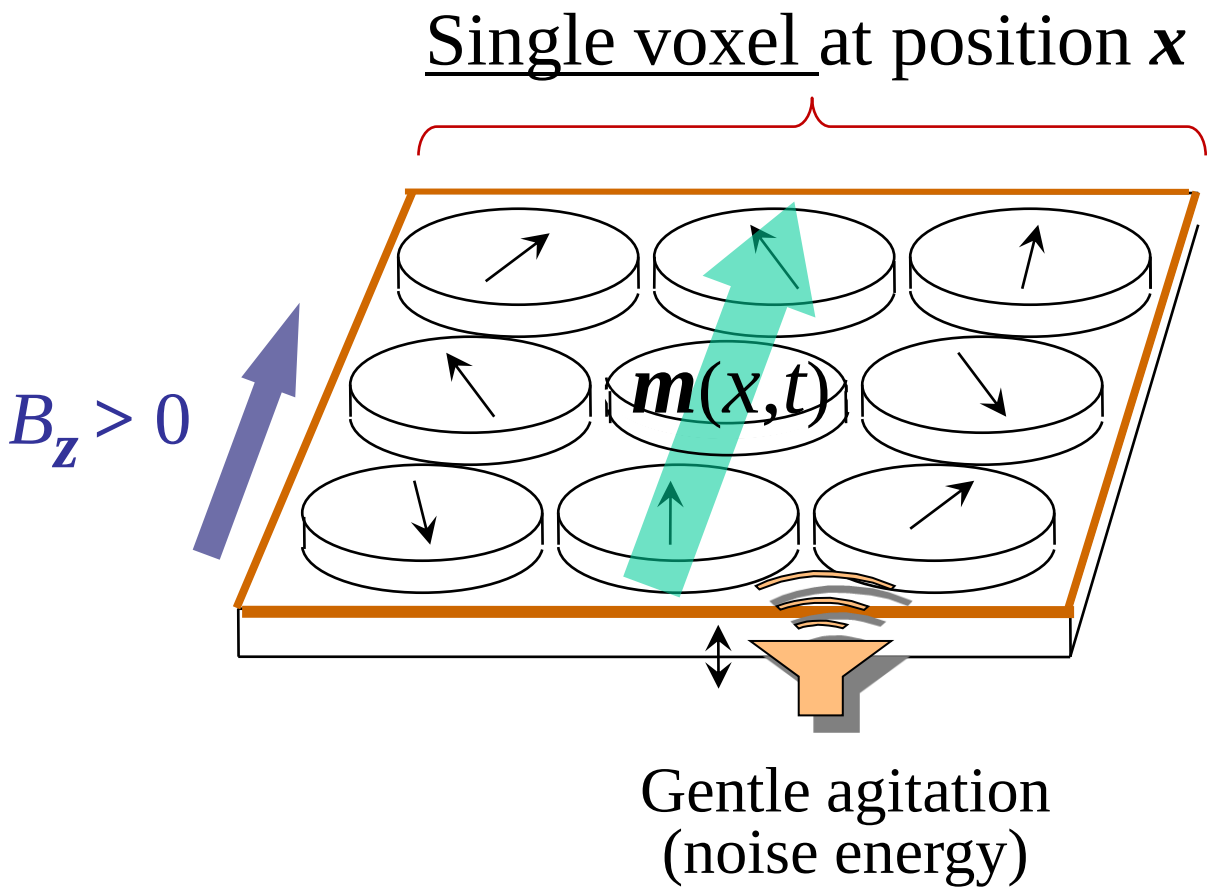


Net magnetization, $\boldsymbol{m}(x,t)$, of the voxel at x
T1 Spin-Relaxation of $\boldsymbol{m}(x,t)$,
T1 MRI of the 1D Patient
Sketch of the MRI Device

Net magnetization, $\boldsymbol{m}(x,t)$, of the voxel at x

Net Magnetization, $\boldsymbol{m}(x,t)$, for the Single Voxel at Position $\boldsymbol{x}$:

magnetic field *from* the ensemble of protons or needles *themselves*



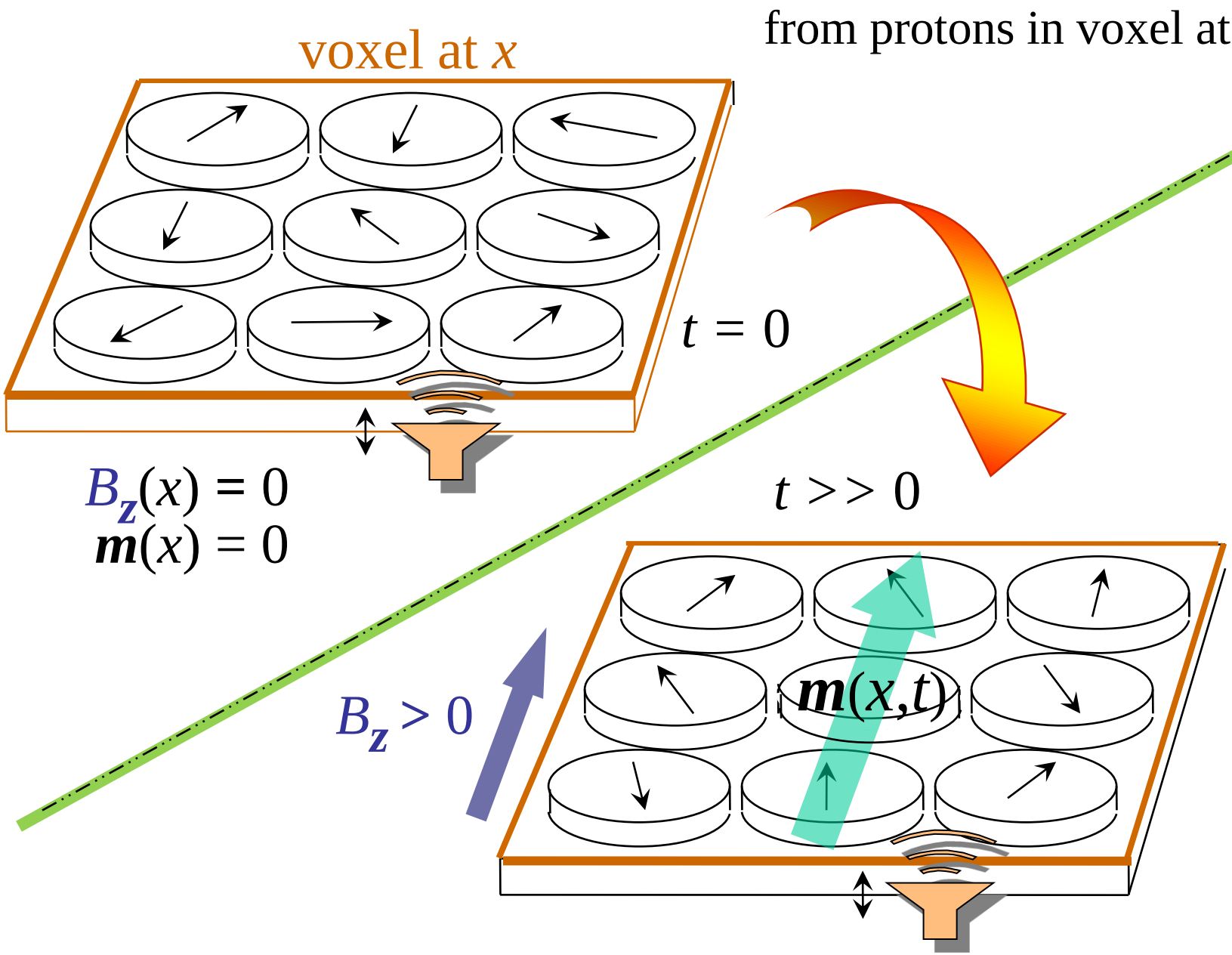
Voxel's MRI Signal Proportional to Its Magnetization, $\boldsymbol{m}(x,t)$

***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?**

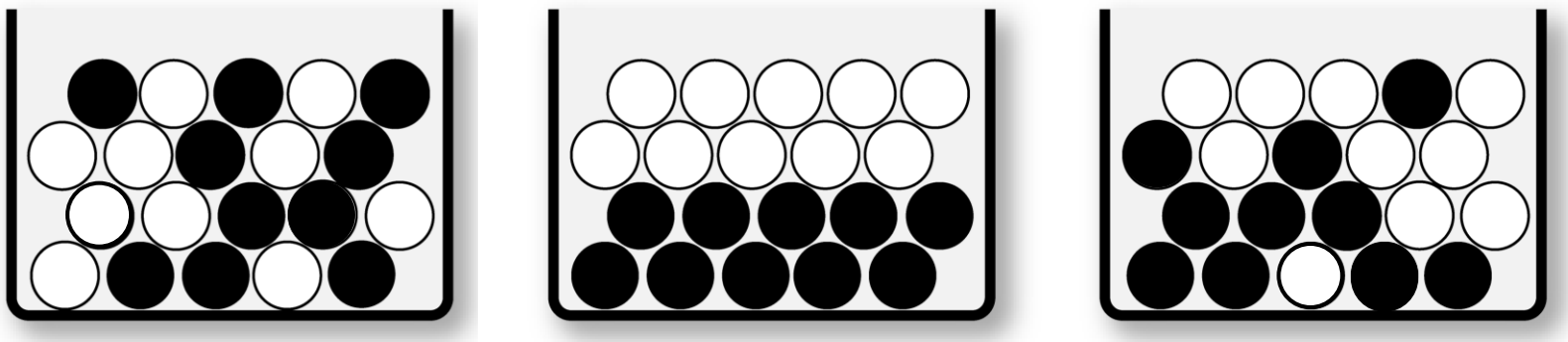
Switching $B_z(x)$ On at $t = 0$ Induces Magnetization $\boldsymbol{m}(x,t)$



Filling Four Energy Levels of Marbles *vs.* Noise Level

equilibrium from battle between energy and entropy

(all slippery; black balls denser)



Shaking (Noise) Energy:

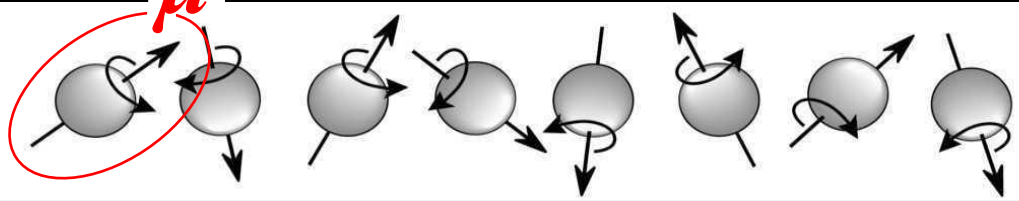
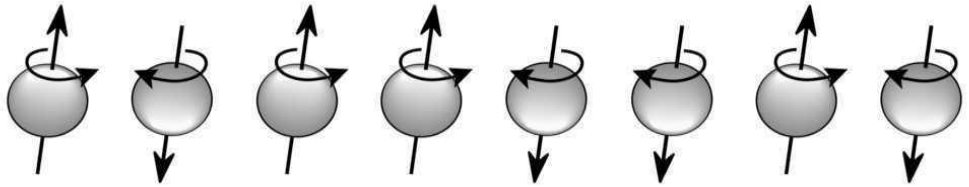
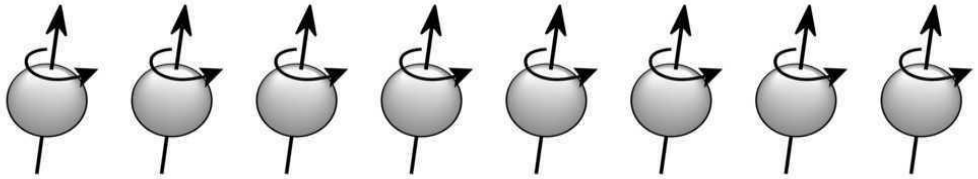
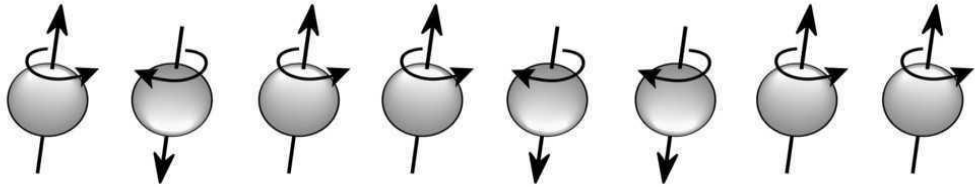
too much *too little* *just right*

Magnetization in Voxel at $\boldsymbol{x}$, under Dynamic Equilibrium:

$$\boldsymbol{m_0}(\boldsymbol{x},t) = [N_-(x,t) - N_+(x,t)]\times\boldsymbol{\mu}\ , \quad t \to \infty$$

collective magnetic field produced by all the protons in it

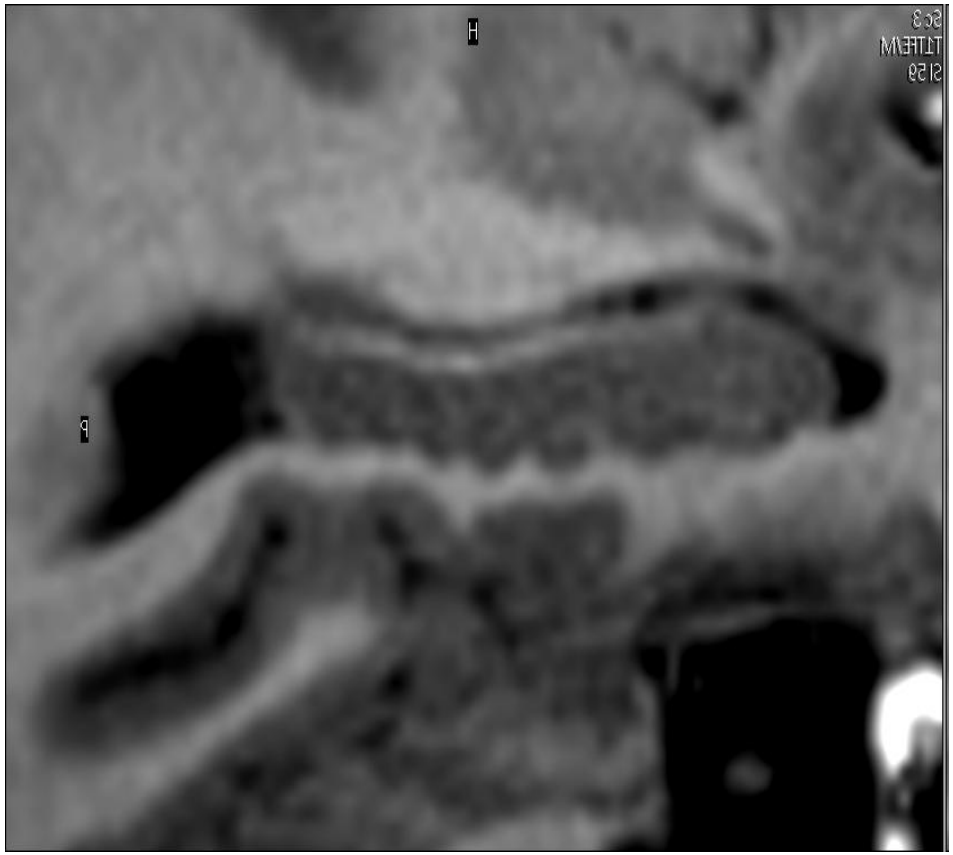
(***Boltzmann***)

B_z	$\boldsymbol{\mu}$	$\boldsymbol{m_0}$
0 tesla		0
0.01 T *or* t = 0+*		~0
»1.5 T		$N\boldsymbol{\mu}_z$
1.5 T $t >> 0$		$5\times10^{-6}\, N\boldsymbol{\mu}_z$

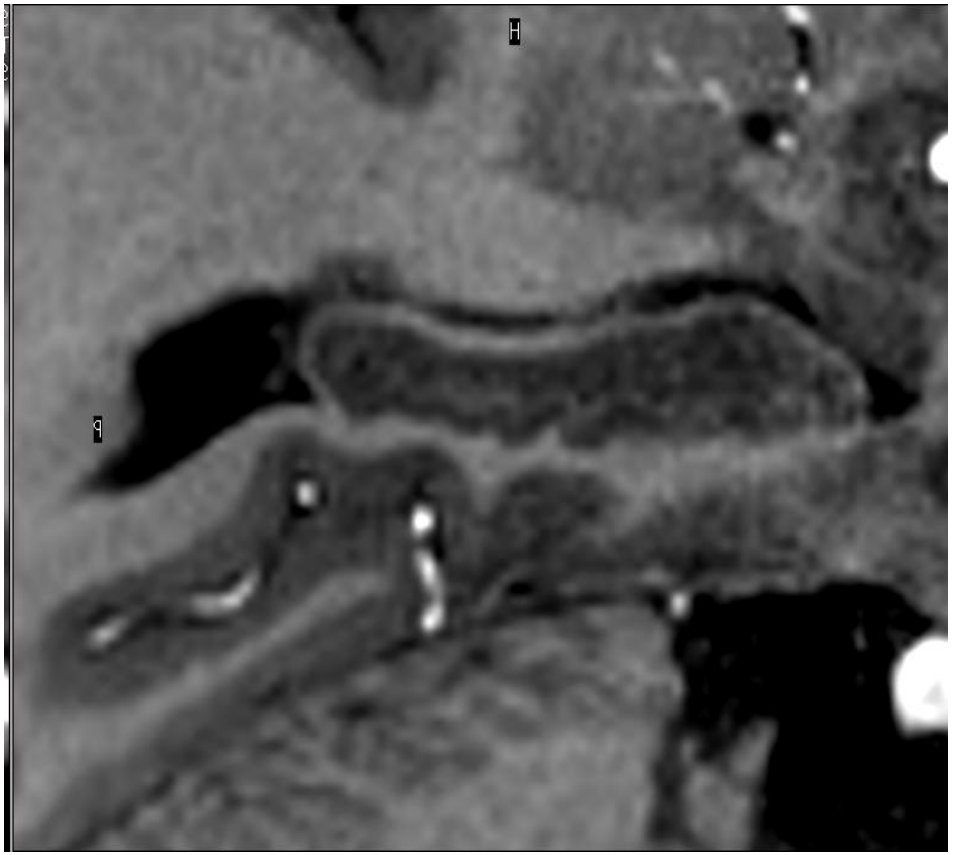
* after abruptly turning on $\boldsymbol{B}_z$, or after a 90° pulse.

$|m_0(x)|$ 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 $\boldsymbol{m}(x,t)$

Voxel's MRI Signal Proportional to Its Magnetization, $\boldsymbol{m}(x,t)$

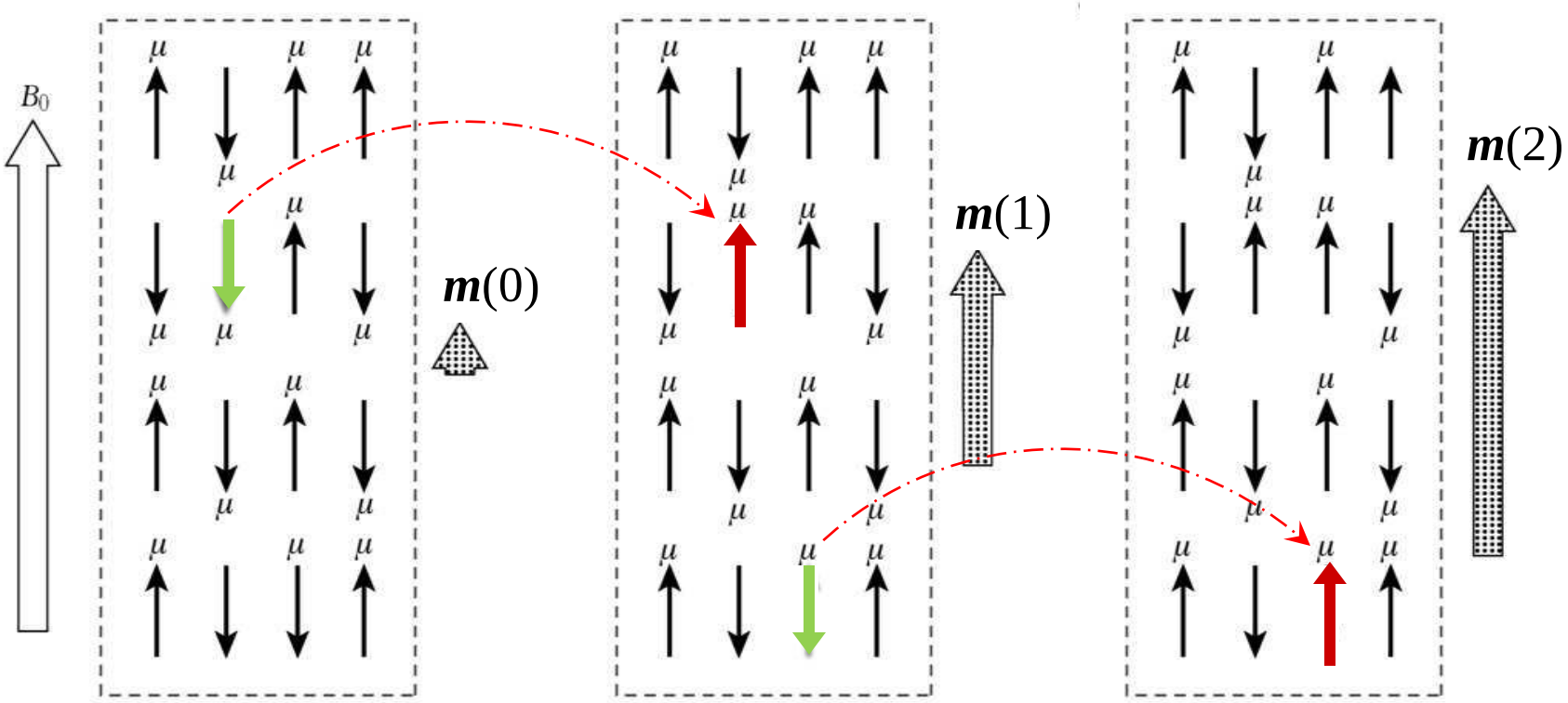
***i*) What is the magnitude of voxel magnetization at thermal equilibrium, m_0?**

***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 ‘down’ spins from ‘outside’



spin ‘tickled’ *down* by f_{Larmor} component of magnetic noise; emits RF photon, phonon. a few can be kicked ‘up’, as well.

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

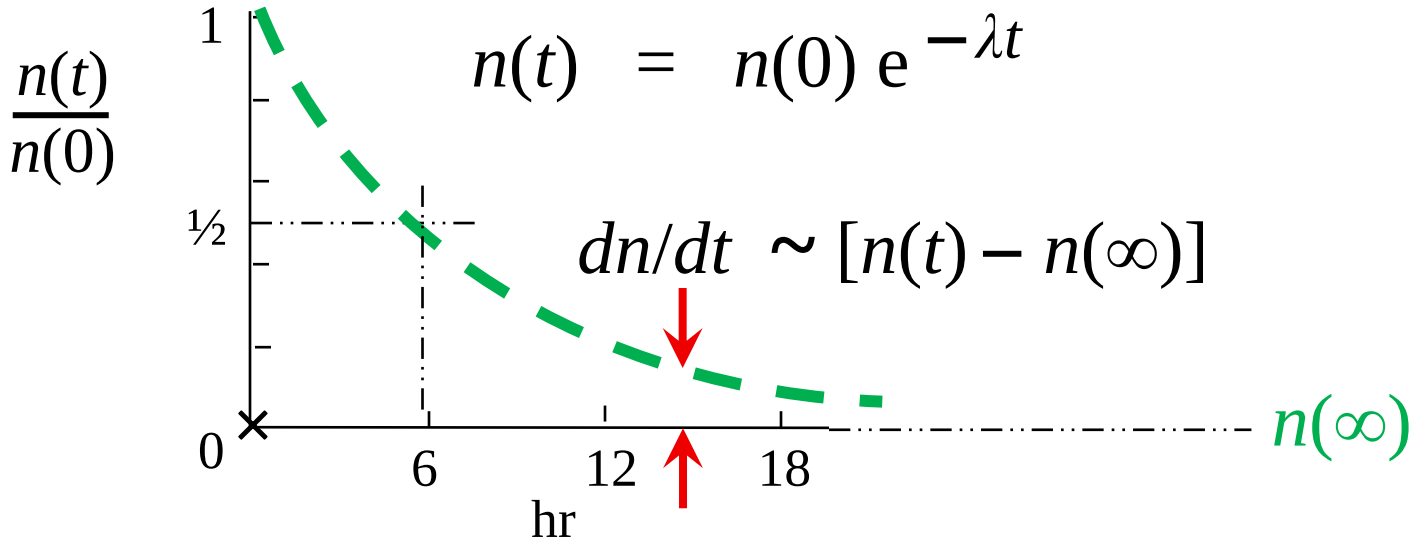
$n(0)$ radionuclei at initially

$n(t)$ remaining after time t.

$n(\infty)$ at 'equilibrium' $n(\infty) = 0$

$$dn/dt(t) = -\lambda\, n(t)$$

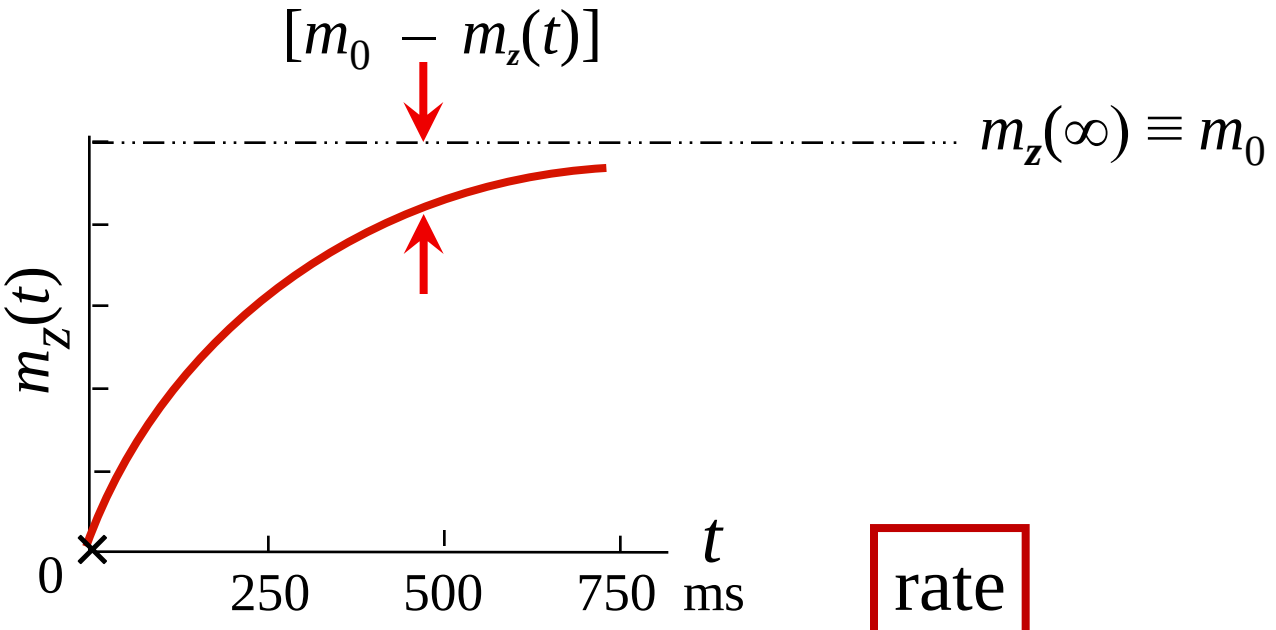
$$n(t) = n(0)\, e^{-\lambda t}$$



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, $m_z(\infty) \equiv m_0$



$$d[m_0 - m_z(t)] / dt = -(1/T1)\,[m_0 - m_z(t)]$$

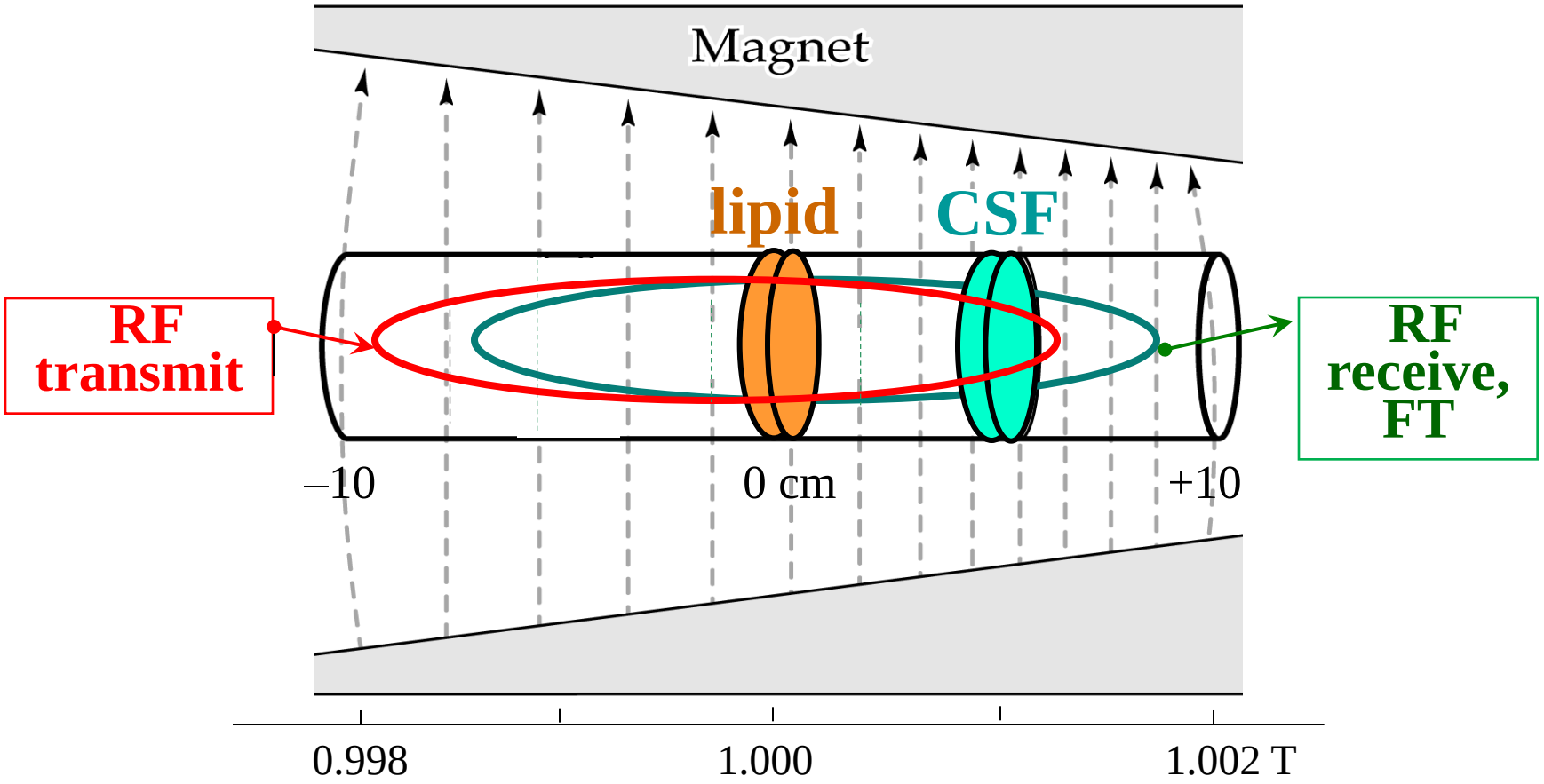
$$m_z(t)/m_0 = 1 - e^{-t/T1}$$

Actually, dynamic equilibrium more complicated: Spins tickled down *and up!*

T1 MRI of the 1D Patient

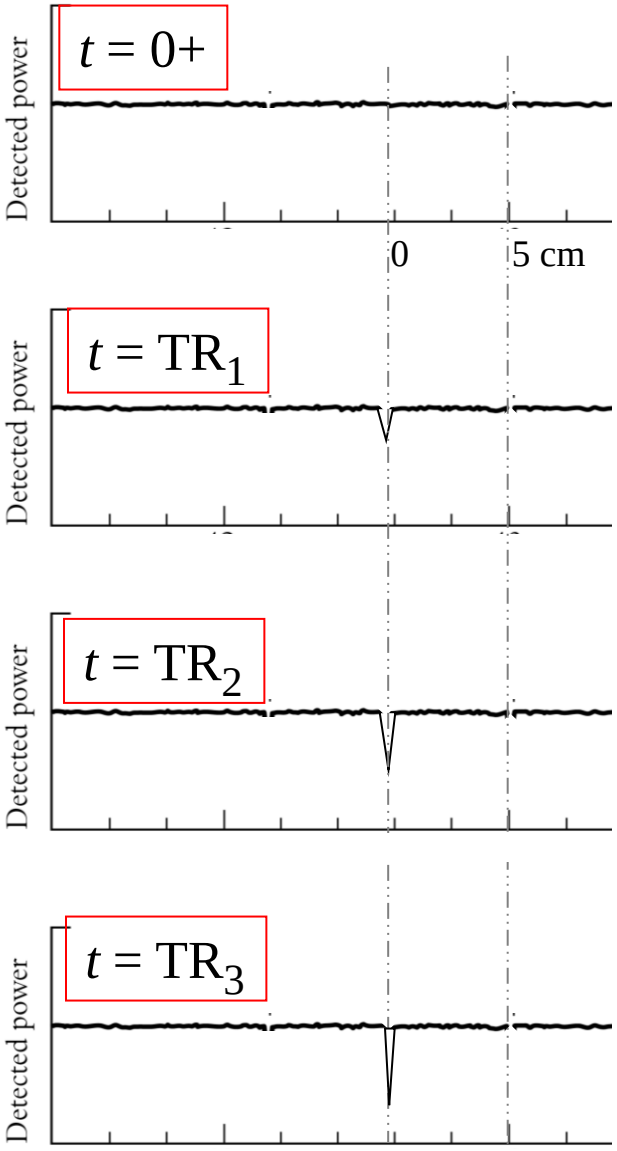
T1 for 2 Voxels: Lipid *vs.* CSF

both at f_{Larmor} of H_2O for local field



Recovery of $\boldsymbol{m}_z(t)$ over Time, from $\boldsymbol{m}_z(\mathrm{TR}_i)$ Measurements

longitudinal component of net voxel magnetization, $\boldsymbol{m}_z(t)$

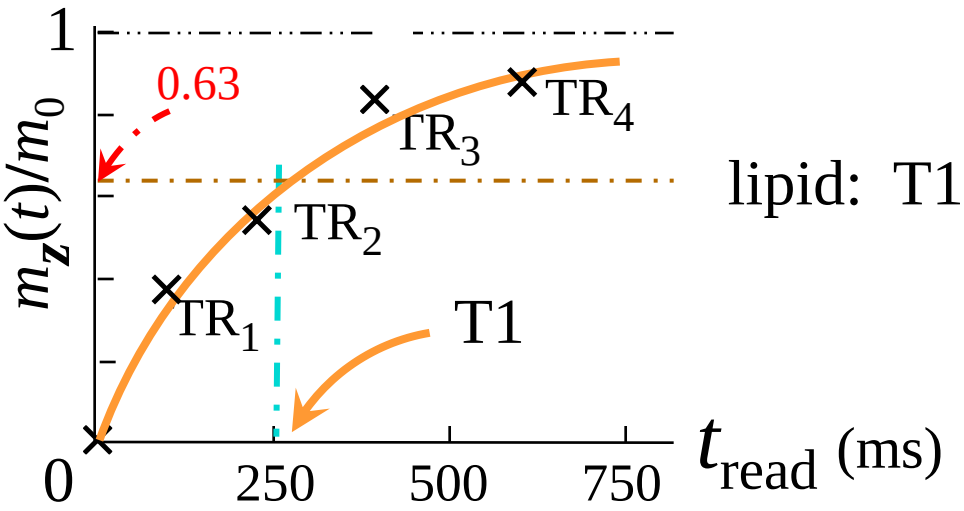


1. First, do ***lipid*** at $x = 0$.
2. Strong burst of RF over range 42.58 MHz ± 250 Hz causes

   $\Rightarrow N_- \sim N_+$
3. After TR_1 delay, sweep through f_{Larmor}, record peak's amplitude.
4. Repeat for several other TR_i values.
5. Plot.

$$m_z(\mathrm{TR})/m_0 \;=\; (1 - e^{-\mathrm{TR}/\mathrm{T1}})$$

curve-fit for set of $\{\mathrm{TR}_i\}$ for voxel at $x = 0$



lipid: T1 ~ 250 ms

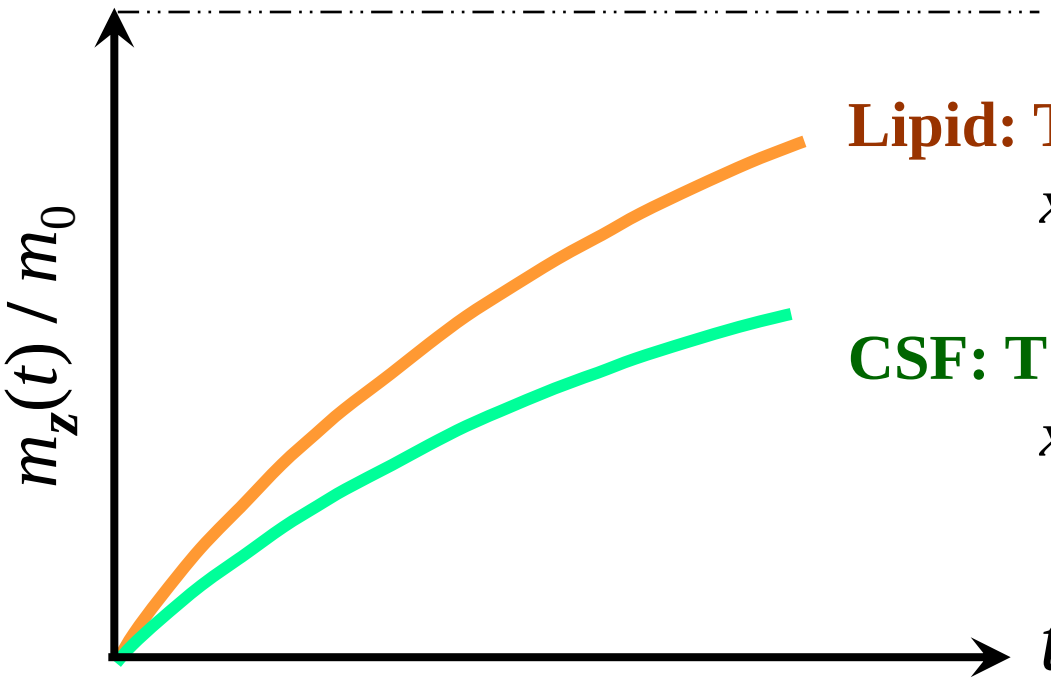
T1

T1 MR image

by convention for T1-imaging: pixel ***white for fast-T1*** tissue
Spritely Brightly

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

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



Lipid: T1 = 250 ms

x = 0

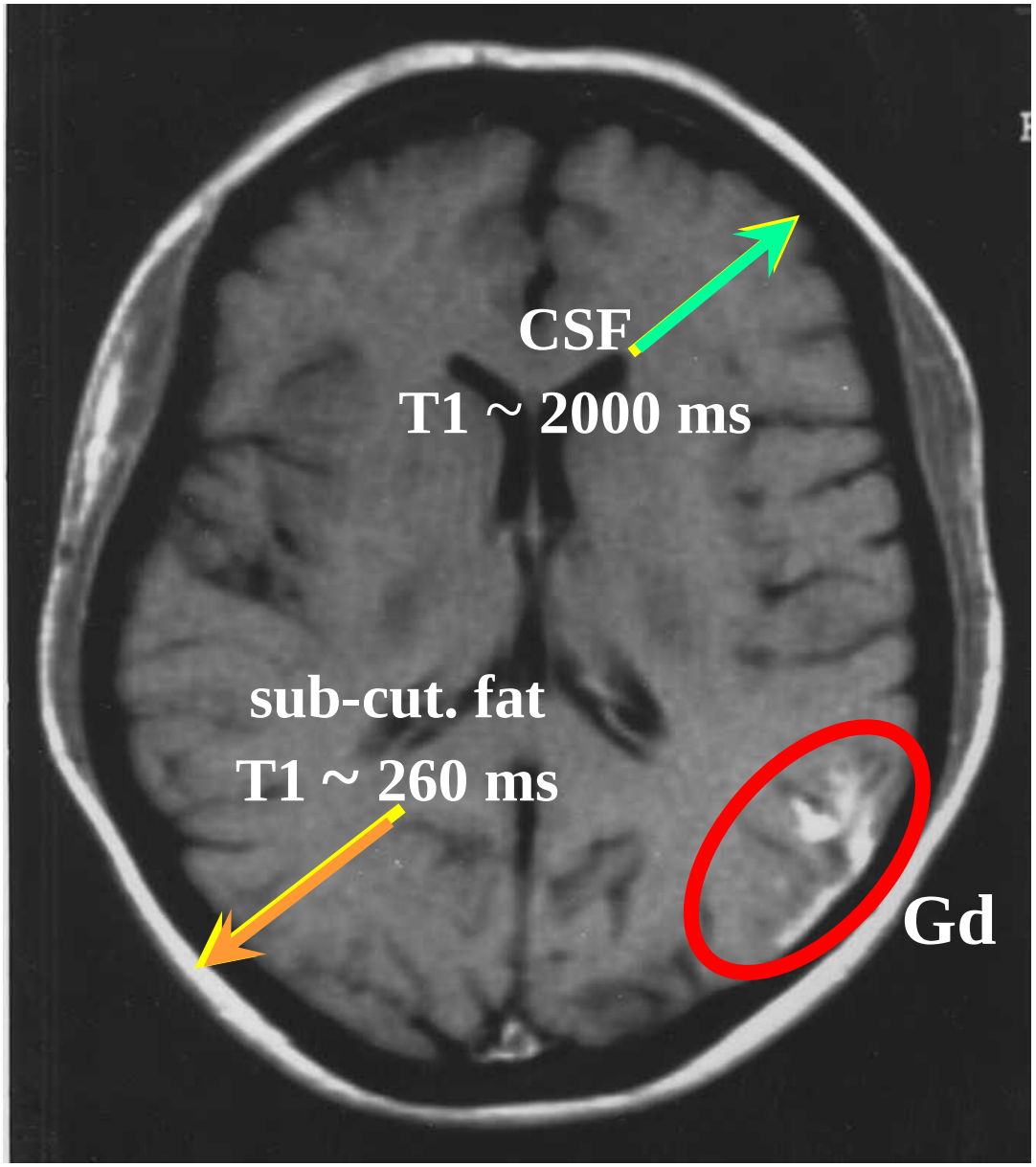
CSF: T1 = 2,000 ms

x = +5 cm

T1 MR image

T1
- lipid: ~ 250 ms
- CSF: ~ 2,000 ms

T1-*w* MR Image



Approximate Relaxation Times of Various Tissues

Tissue	PD p^+/mm^3, rel.	T1, *1 T* (ms)	T1, *1.5 T* (ms)	T1, *3 T* (ms)	T2 (ms)
pure H_20	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, $\boldsymbol{m}(x,t)$

***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 ***only*** thing a proton is ***ever*** aware of, or reacts to, is the ***local*** *magnetic field*, $\boldsymbol{B}_{\text{local}}(t)$.

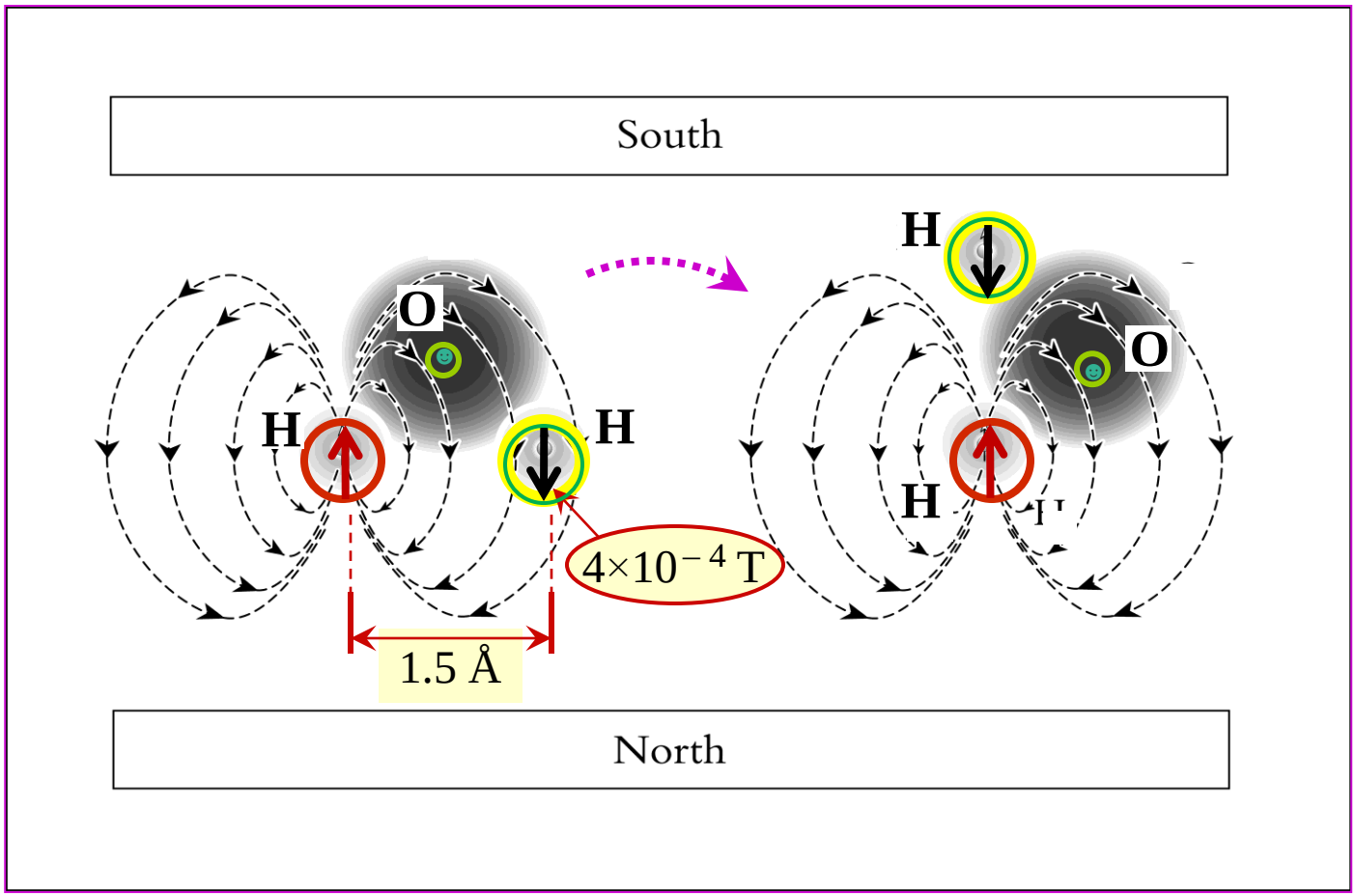
!!!

But the source of $f_{\text{Larmor}}(\boldsymbol{B}_{\text{local}}(t))$ can be either ***external*** ($\boldsymbol{B}_{\text{RF}}$) or ***internal*** (*e.g.*, moving partner)

!!!!!

Random $B_{RF}(t)$ at f_{Larmor} Causes T1-Transitions

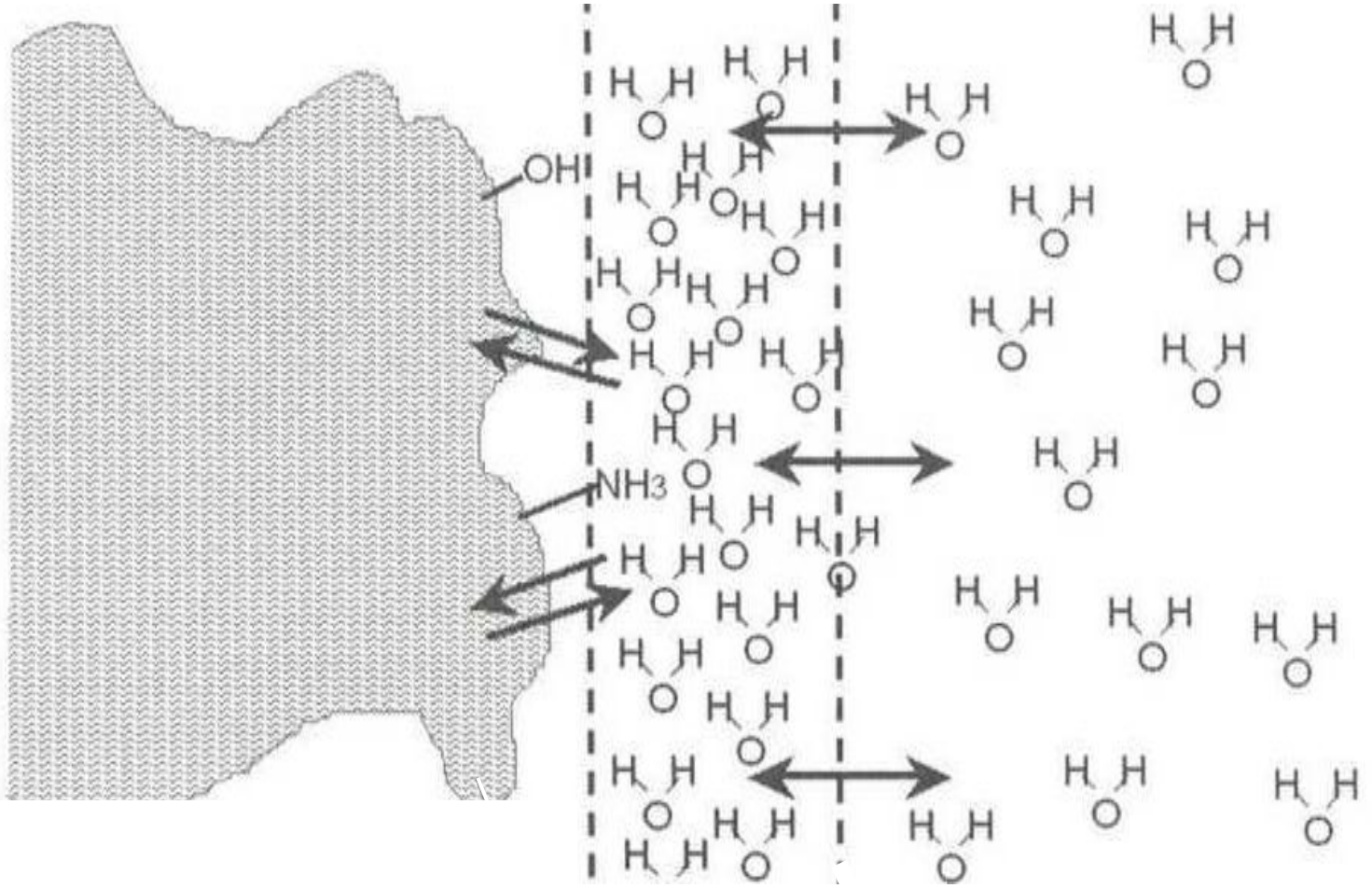
variations in proton magnetic dipole-dipole interactions



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

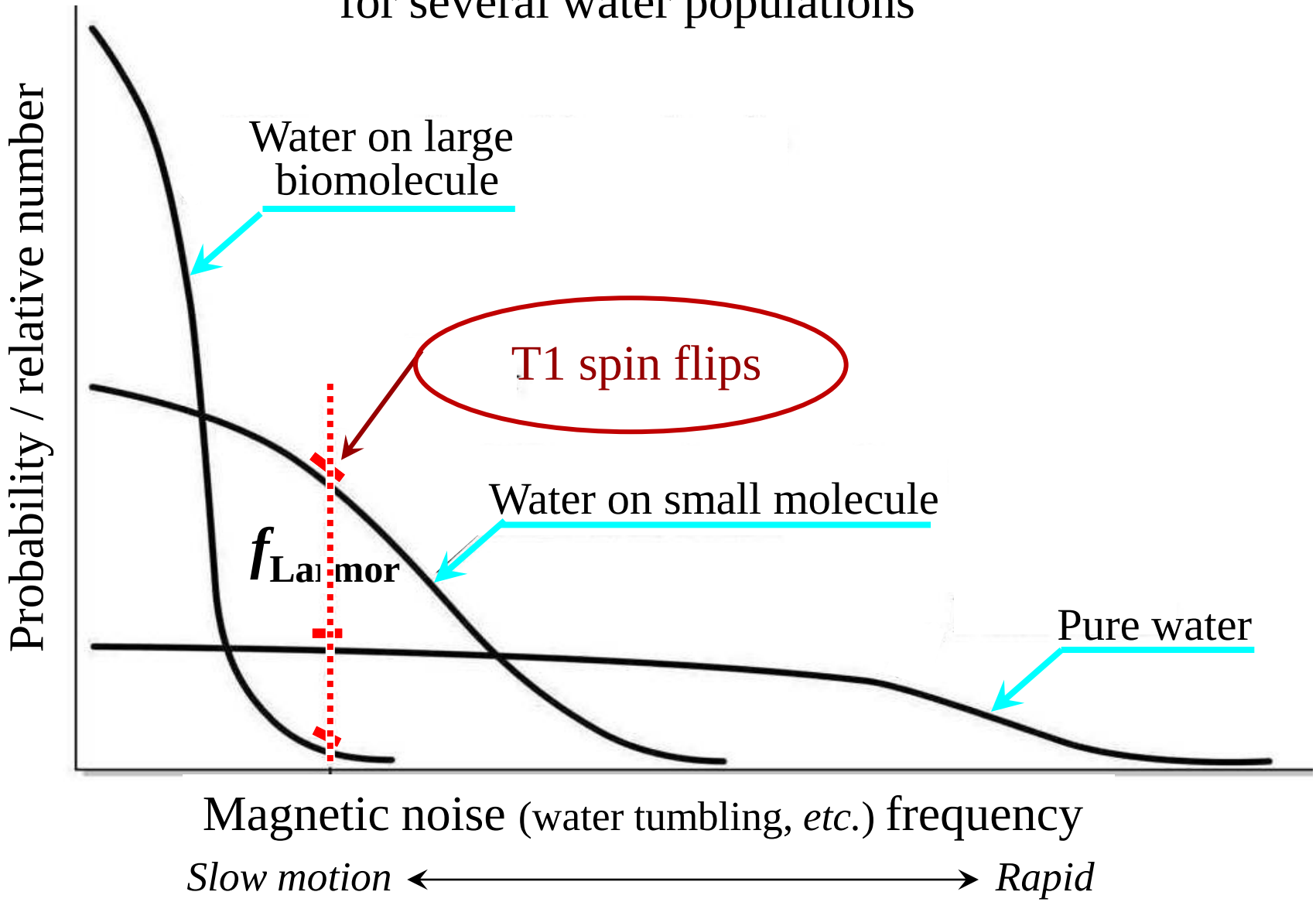
Restrictions on Rotations of Water Molecules

more or less bound *hydration layer* *free*

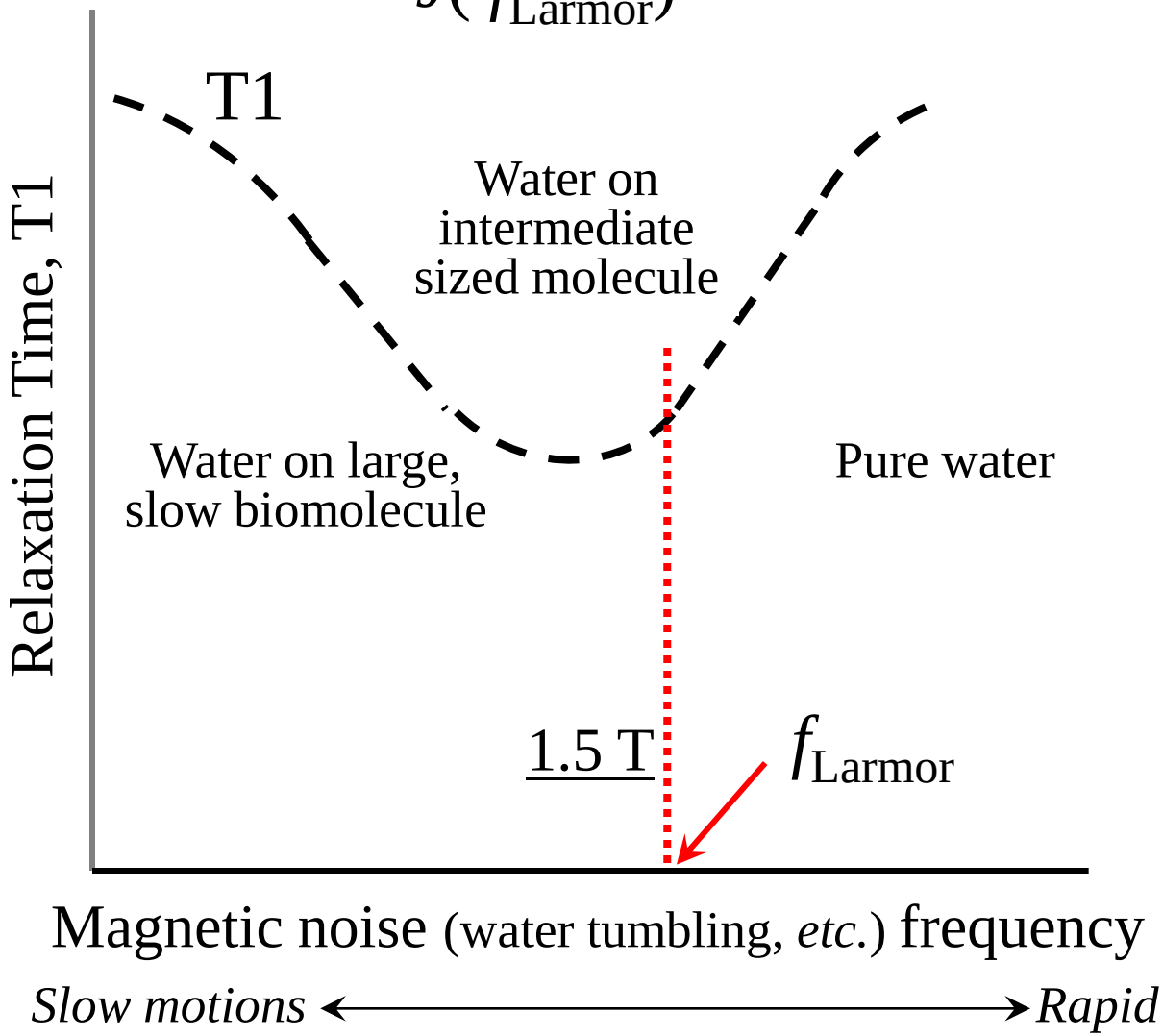


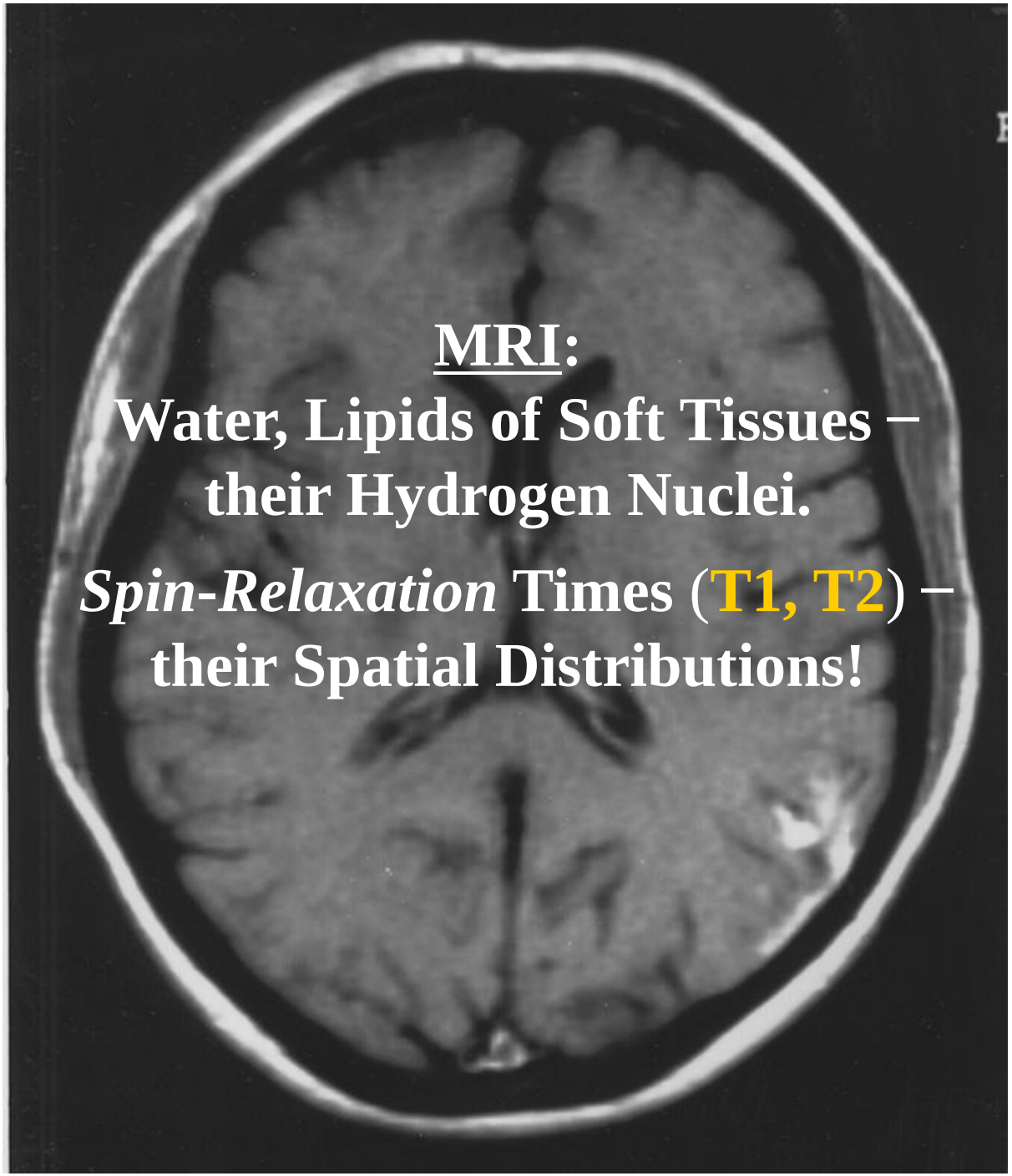
Rotational Power Spectra, $J(f)$, of Water

for several water populations



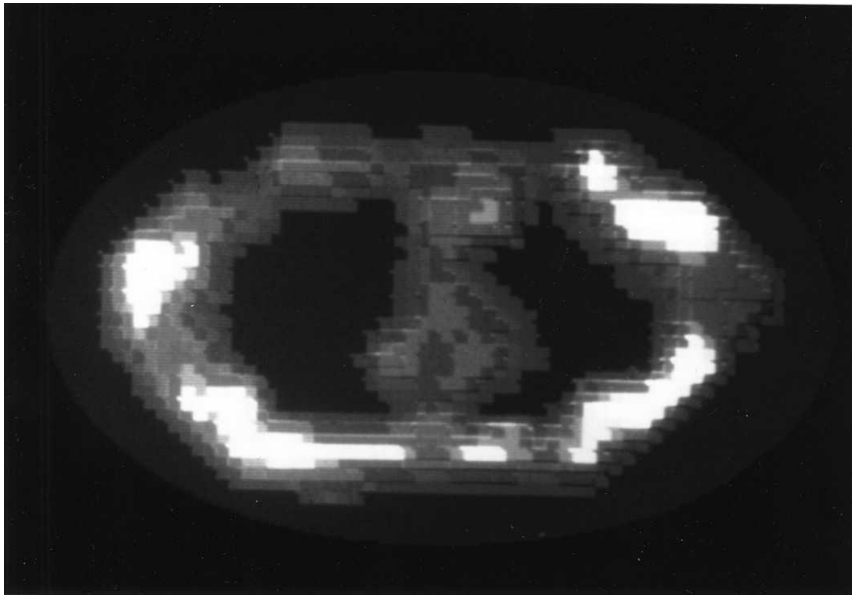
Average Water T1 Determined by Noise Power Spectrum, $J(f_{Larmor})$




MRI:
Water, Lipids of Soft Tissues –
their Hydrogen Nuclei.
Spin-Relaxation Times (T1, T2) –
their Spatial Distributions!

Sketch of the MRI Device

Damadian's Indominable, 1977 (Smithsonian)

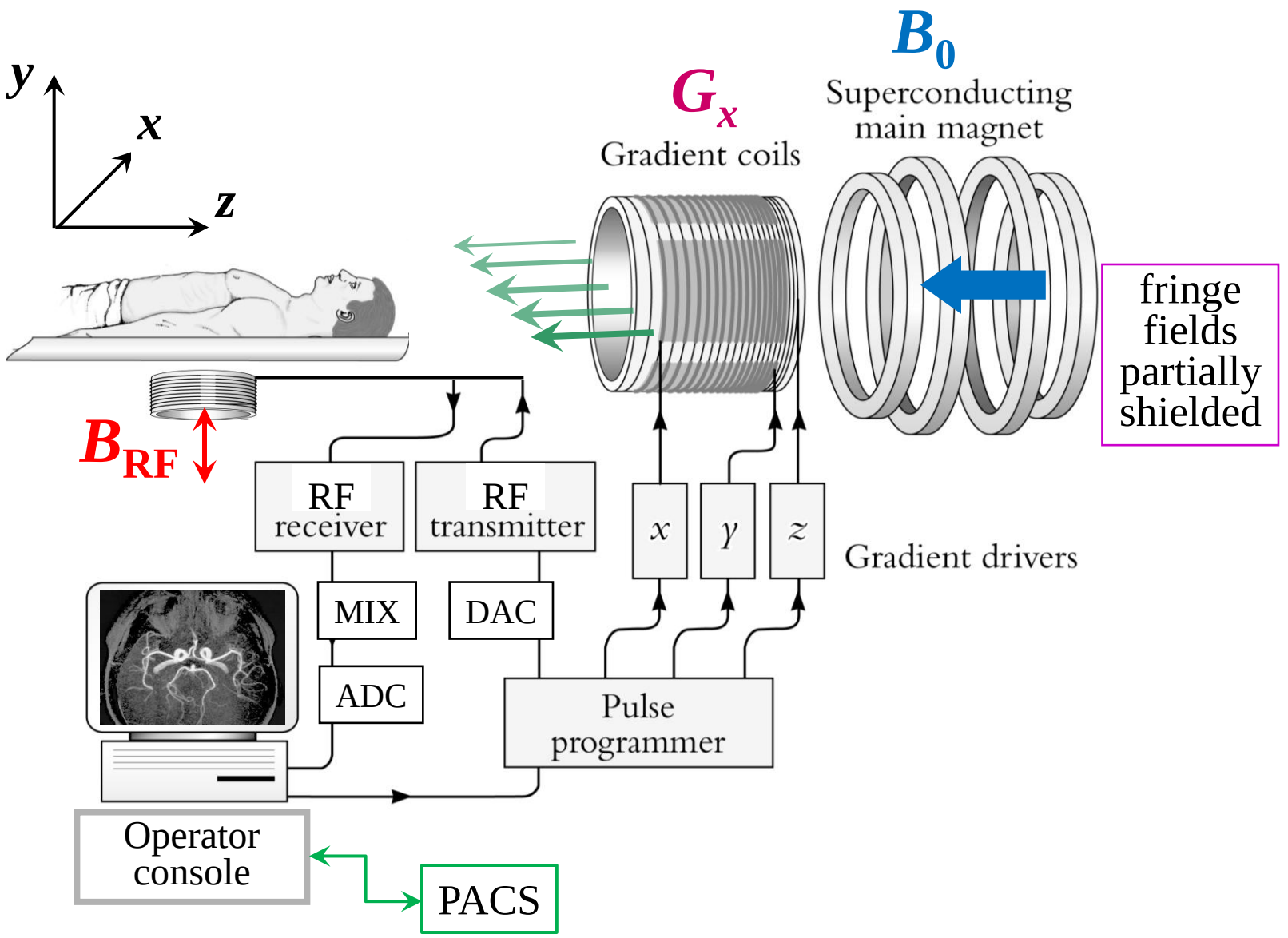


Nobel Prize, 2003

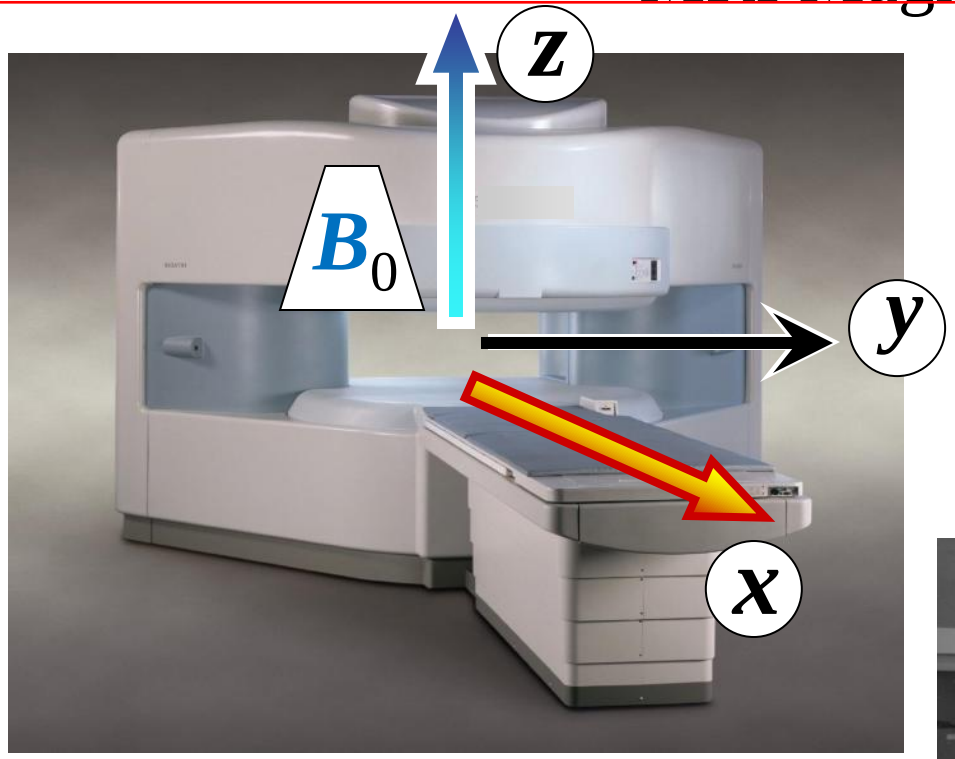
Paul Lauterbur

Peter Mansfield

Major Components of a *Superconducting* MRI System

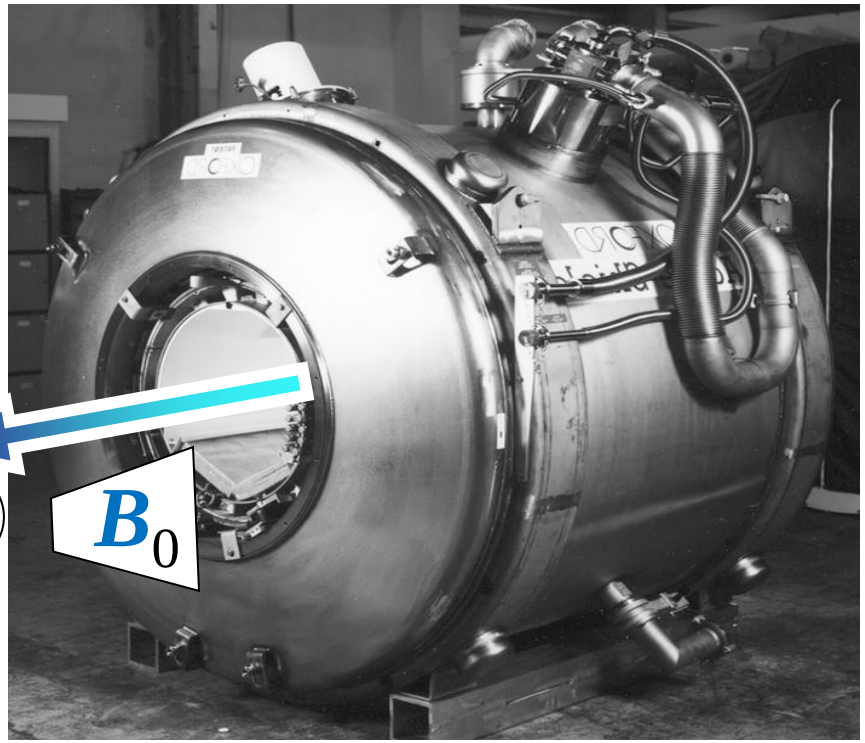


MRI Magnets



Open (*most of this presentation*)
electromagnetic or permanent

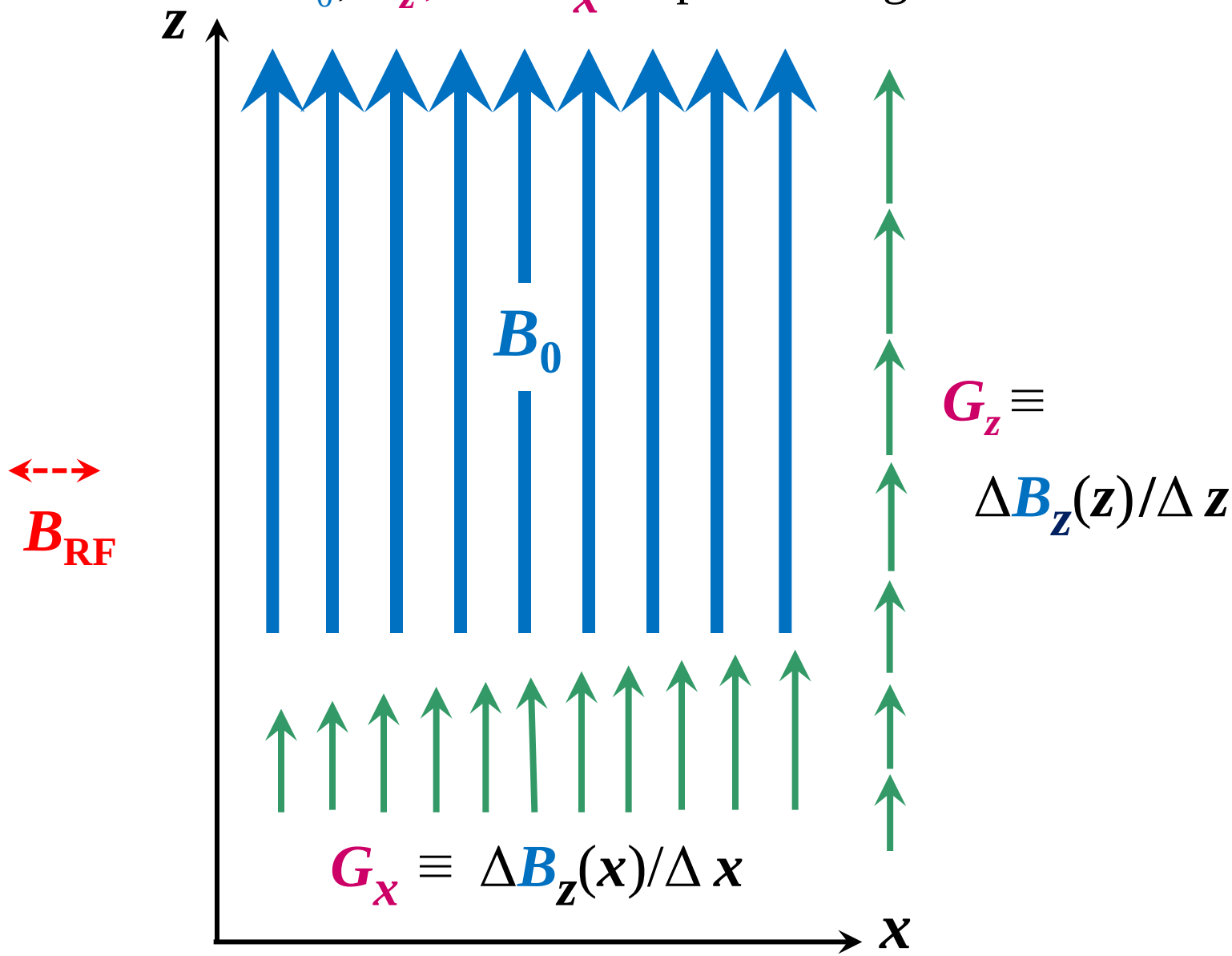
Superconducting
e.g., niobium-titanium wire



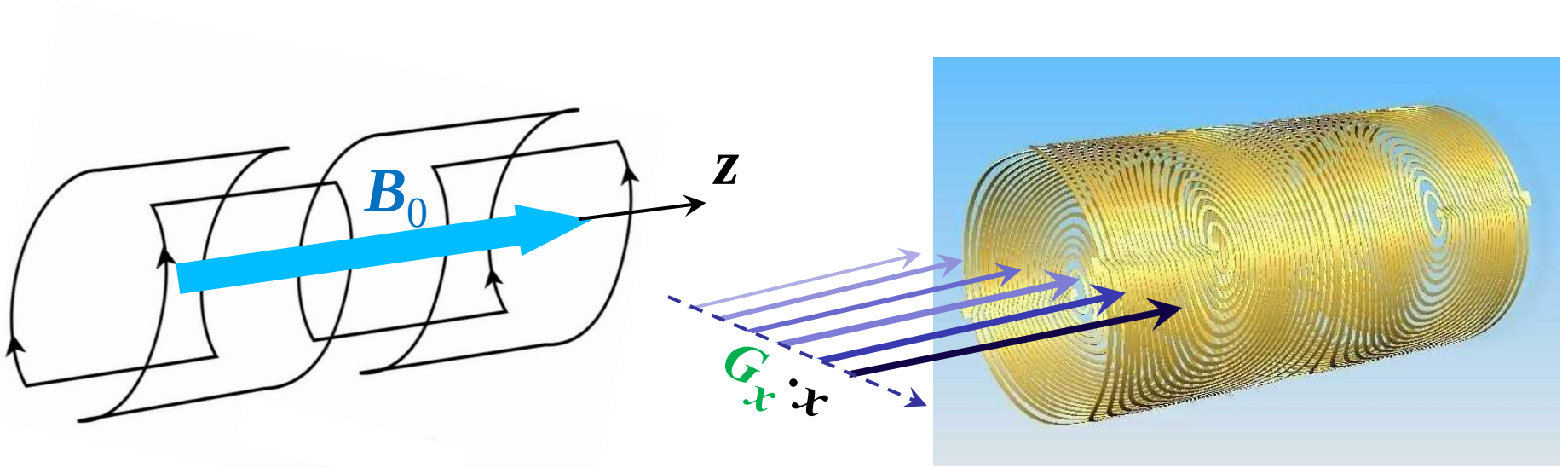
B_0 : < 0.5T, 1.5 T, 3.0 T, (7 T)
Homogeneity: <10 ppm
Shielding: passive and active
Cryogen: 0.1 liter He/y
Weight: 4 tons (supercond.)

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

B_0, G_z, and G_x all point along z!



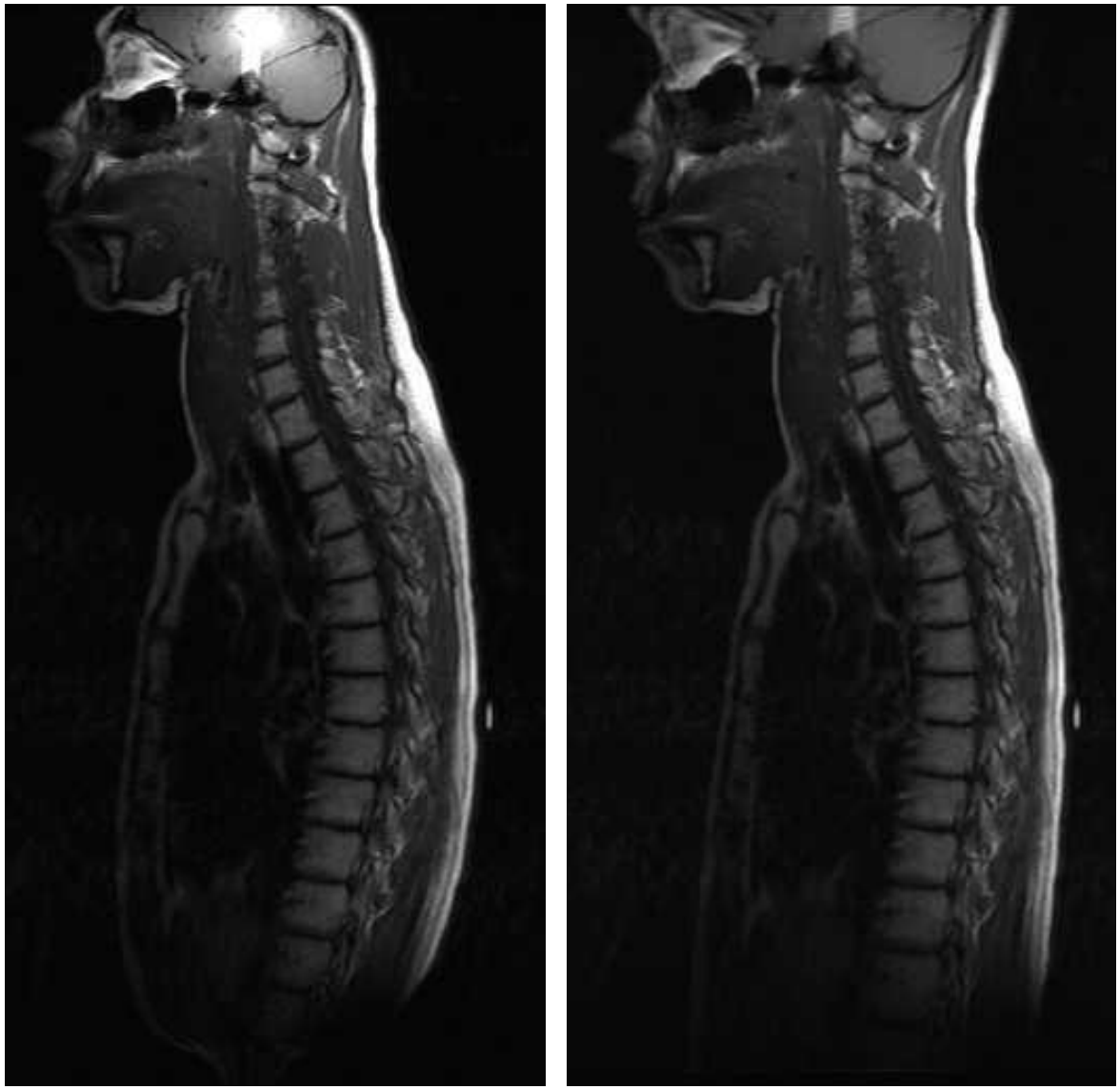
x-Gradient Magnet Winding for Superconducting Magnet



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

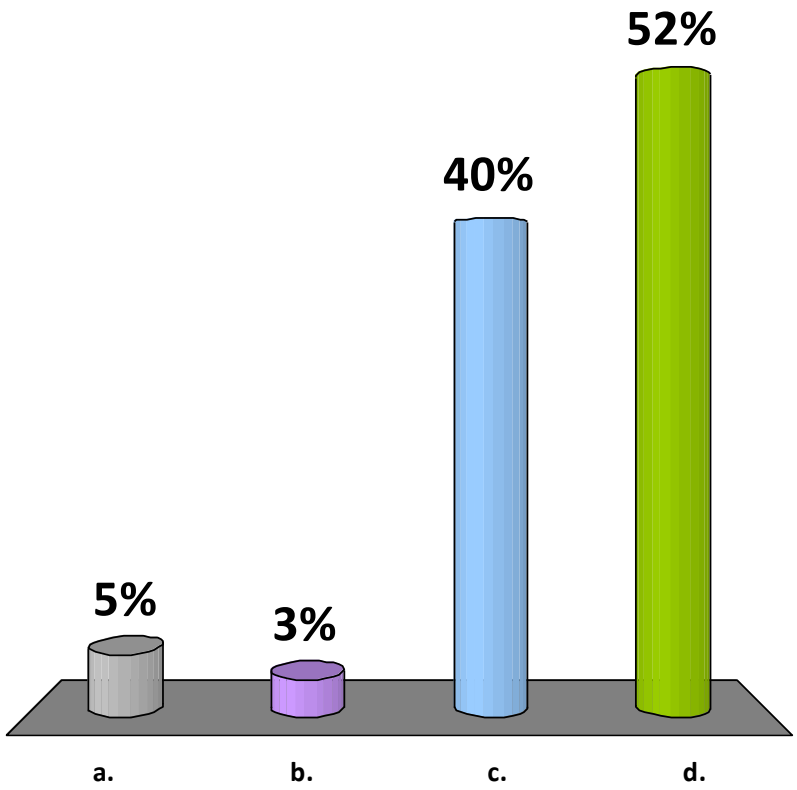
Artifact: Gradient Non-Linearity



SAMS Question

II-4 Gradient coils affect the strength of the magnetic field that, at all locations in space, point…:

a. along the x direction of the scanner
b. along the *y* direction of the scanner
c. along the *z* direction of the scanner
d. along all directions of the scanner
not enough information given here to answer



SAMs Q:

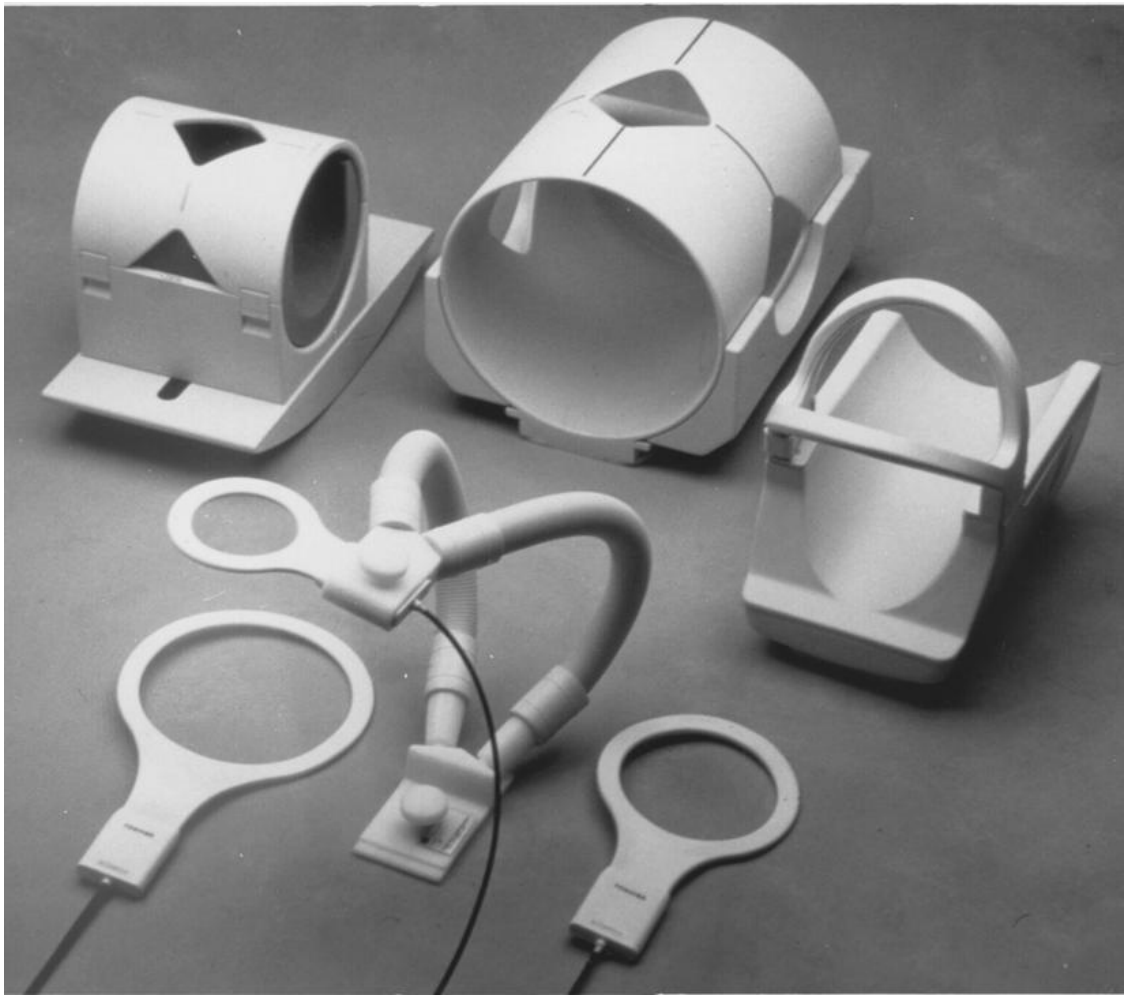
II-4 Gradient coils affect the strength of the magnetic field that, at all locations in space, point…:

(a) along the *x* direction of the scanner
(b) along the *y* direction of the scanner
(c) along the *z* direction of the scanner
(d) along all directions of the scanner
(e) not enough information given here to answer

Answer: (c). Along the principal magnetic field, B_0, hence the *z*-axis

Ref: "Medical Imaging", A.B. Wolbarst *et al.*, Wiley-Blackwell (2013), p. 316.

RF Coils



B_{RF} : 20 μT
Pulse on-time: 3 msec
RF power: 15 – 25 kW
SAR: 2 – 20 W/ kg

‘Parallel’ RF Receiving Coils for Much Faster Imaging

transmit coils coming

SAMS Question

II-5 The acronym used to describe the amount of RF radiation uptake in patient tissue is:

4%	a. REM
90%	b. SAR
1%	c. CTDI
3%	d. STIR
2%	e. RARE

SAMs Q:

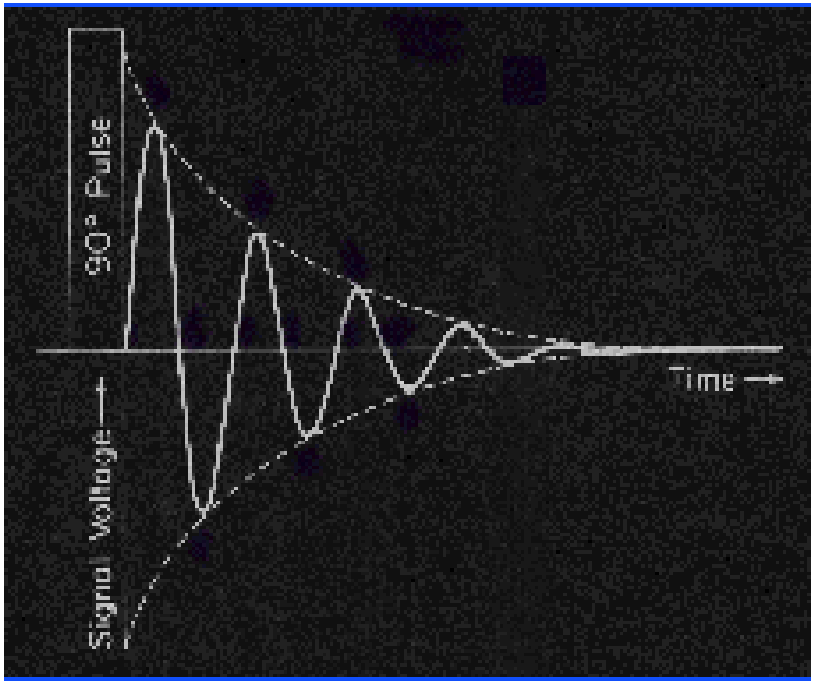
II-5. The acronym used to describe the amount of RF radiation uptake in patient tissue is:

(a) REM
(b) SAR
(c) CTDI
(d) STIR
(e) RARE

Answer: (b). Specific Absorption Rate

Ref: "Medical Imaging", A.B. Wolbarst *et al.*, Wiley-Blackwell (2013), p. 349.

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

$|\psi\rangle$

now this

Simple QM

$|\uparrow\rangle, |\downarrow\rangle$

transitions between
spin-up, spin-down states

f_{Larmor} , m_0 , T1

oversimplified

Classical Bloch Eqs.

for expectation values

precession, nutation of
voxel magnetization, $\boldsymbol{m}(t)$

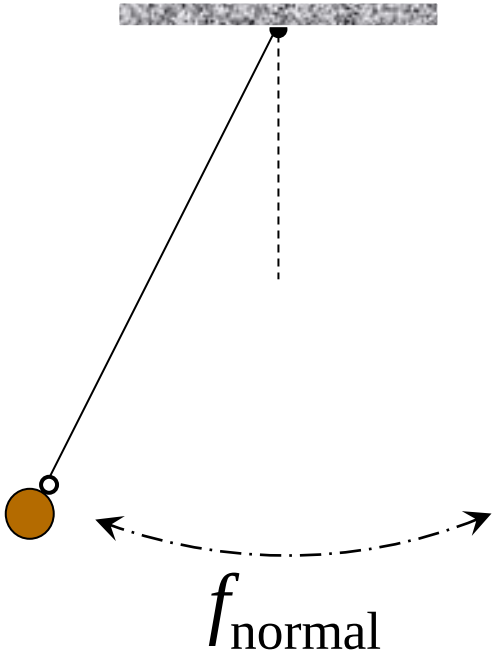
f_{Larmor} , T2 , $\boldsymbol{k}$-space

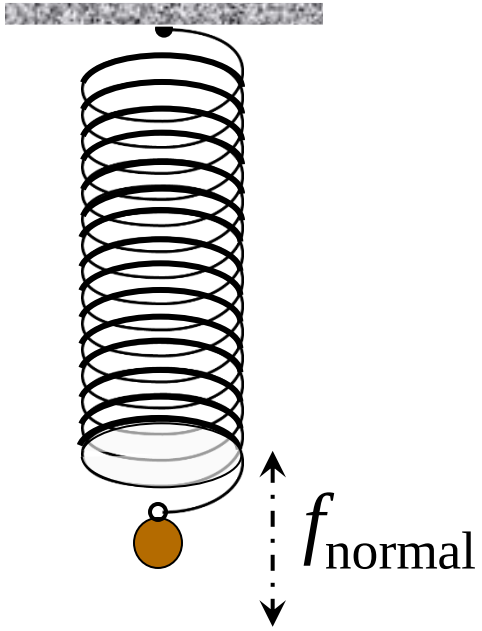
exact; from full QM

Normal Modes, Resonance, Precession, Nutation

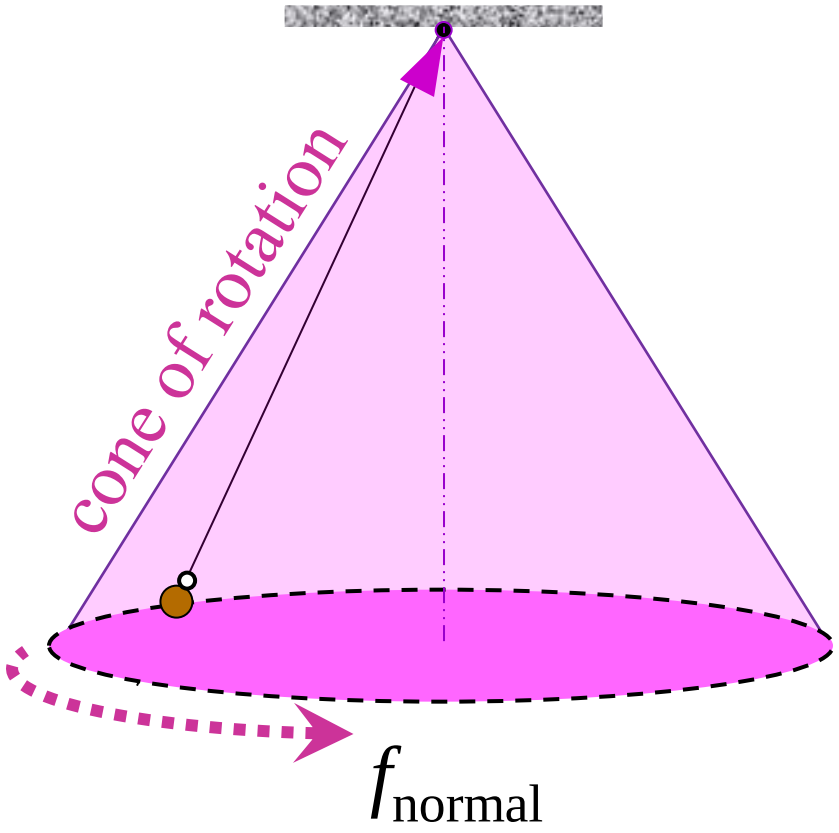
Normal Mode, at f_{normal}

f_{normal}





A Normal Mode of a 2-D Pendulum



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

$J(t)$: Angular Momentum

With torque, τ:

$$dJ/dt = \tau$$

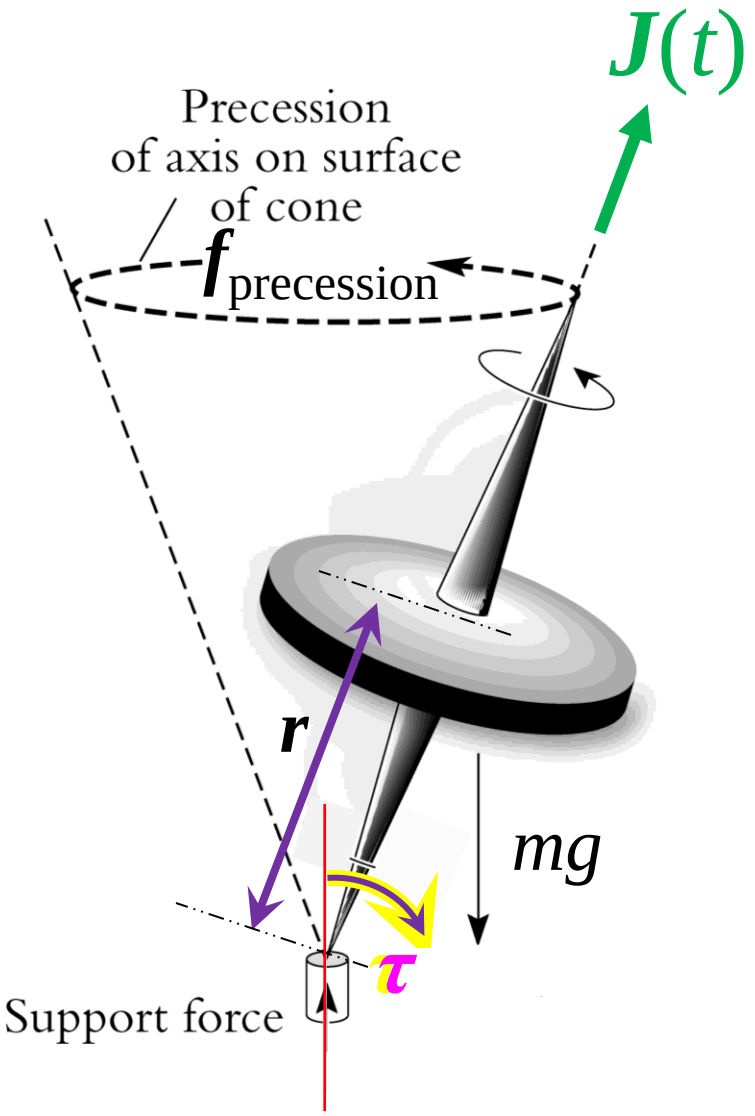
Equation of Motion

Angular acceleration is just like

$$dp/dt = F$$

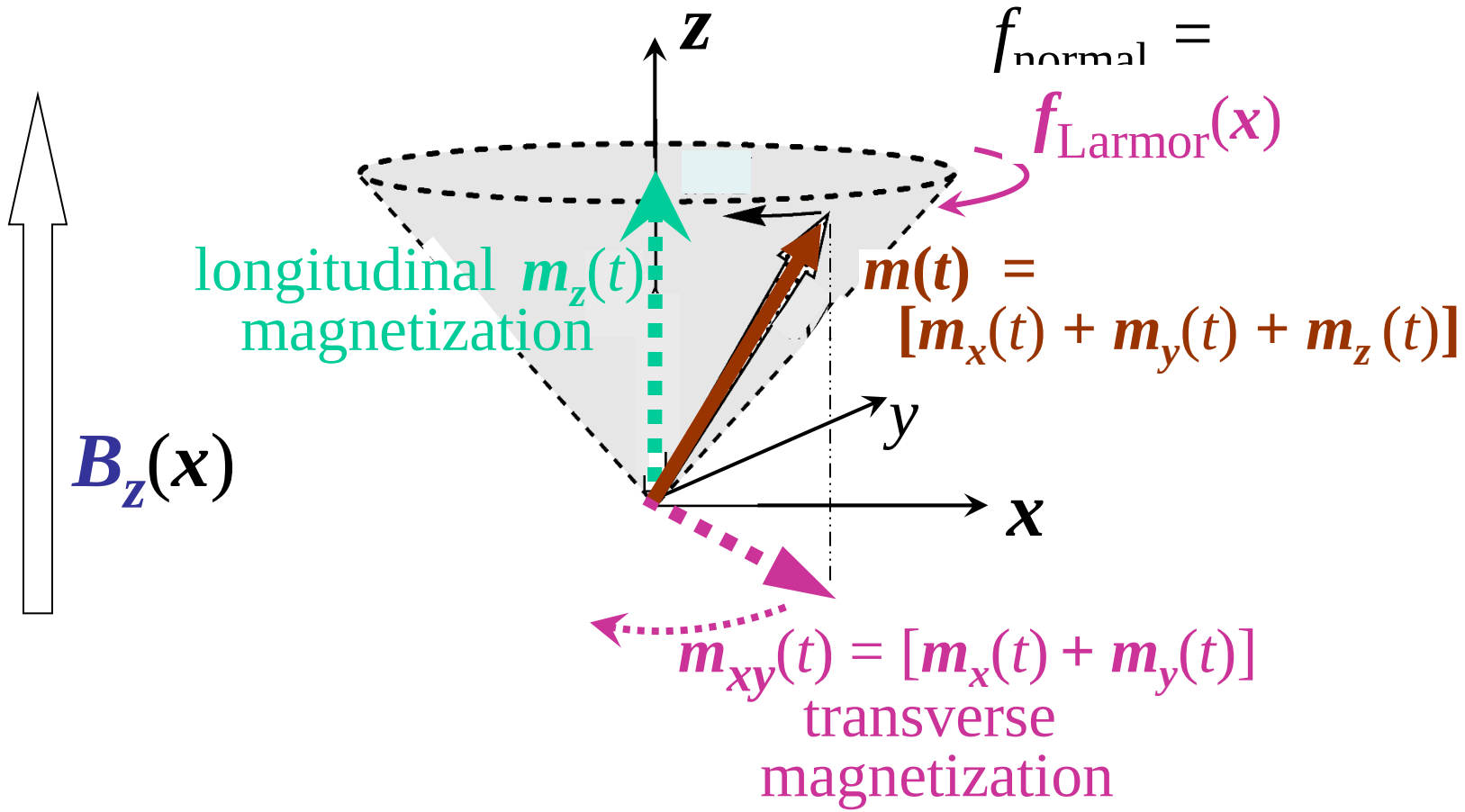
but here, $J(t)$ changes ***direction only:***

Precession at f_{normal}



Normal Mode Precession of Voxel $\boldsymbol{m}(\boldsymbol{x},t)$ in Magnetic Field

can be derived rigorously from quantum mechanics



Classical Bloch Equations of Motion for $m(\boldsymbol{x},t)$ in $\boldsymbol{B_0}$

Equation of motion for spinning body

$$dJ/dt = \tau \quad \text{(external torque)}$$

recall: $\boldsymbol{\mu} = \gamma J$

$$d(\boldsymbol{\mu}/\gamma)/dt = \boldsymbol{\tau}$$

Lorentz torque on spins at $\boldsymbol{x}$ with magnetic moment $\boldsymbol{\mu}$ in $\boldsymbol{B}_z(x)$:

$$\boldsymbol{\tau} = \boldsymbol{\mu} \times \boldsymbol{B}_z(x) \quad (\textbf{\textit{vector}} \text{ cross product})$$

Equation of motion becomes:

$$d\boldsymbol{\mu}(x,t)/dt = \gamma\, \boldsymbol{\mu}(x,t) \times \boldsymbol{B}_z(x) \ .$$

Sum/average over all protons in voxel:

$$d\langle \boldsymbol{m}(x,t) \rangle / dt = \gamma \langle \boldsymbol{m}(x,t) \rangle \times \boldsymbol{B}_z(x)$$

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

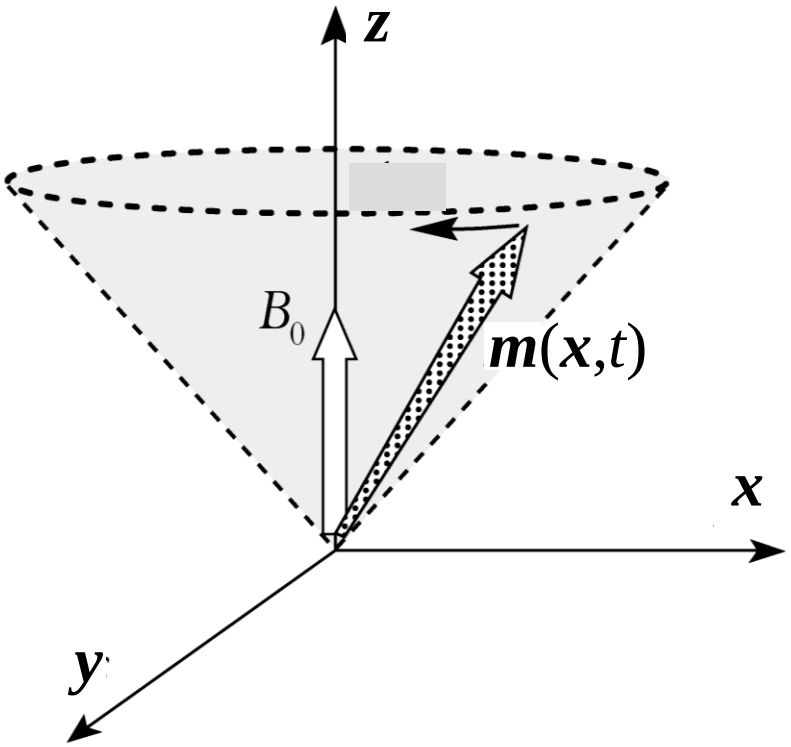
With T1 relaxation along $\boldsymbol{z}$-axis:

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

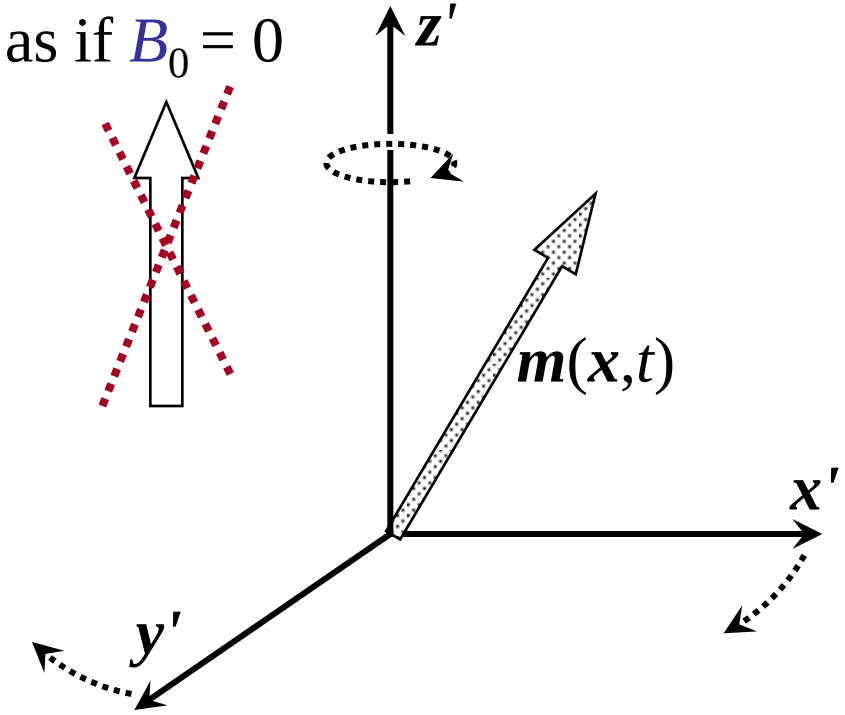
Precession of $\boldsymbol{m}(x,t)$ about $\boldsymbol{B_0}$ at $f_{\text{Larmor}} = (\gamma/2\pi)\, B_z(x)$

as seen from a frame that is

Fixed

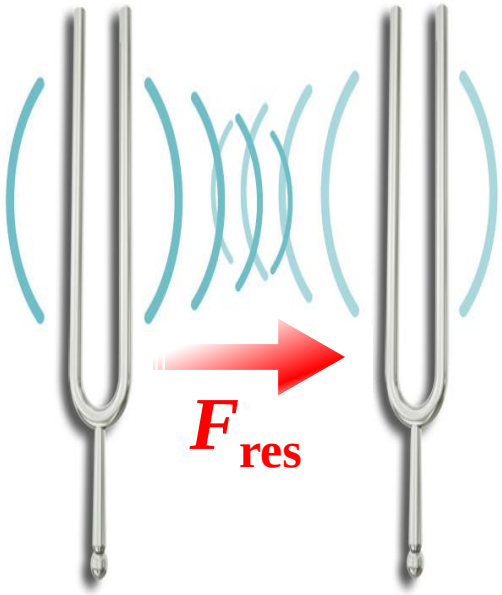


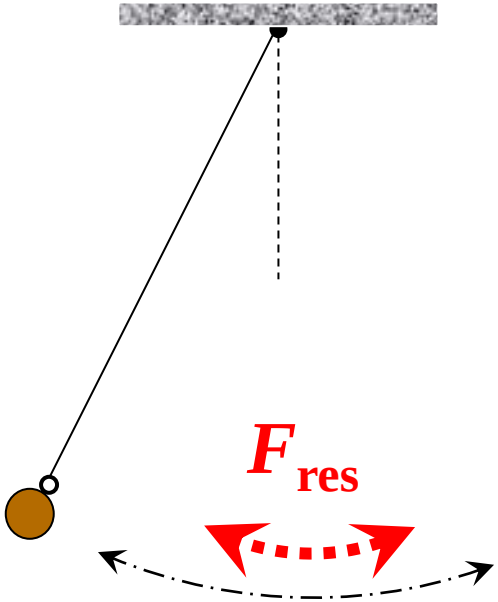
Rotating at $f_{\text{Larmor}}(x)$

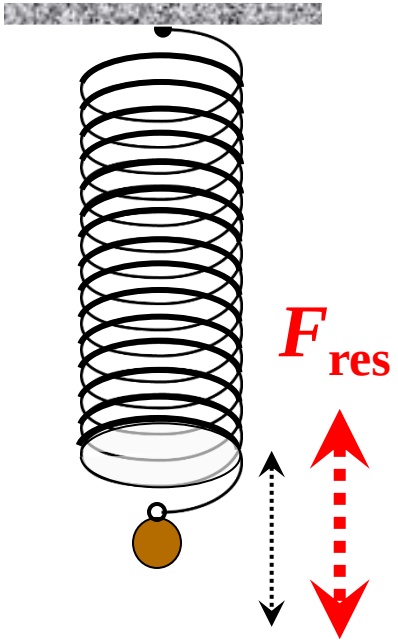


The ponies don't advance when you're *on* the carousel;
It's as if $B_0 = 0$!

Resonance Energy Transfer when $f_{driving} = f_{normal}$

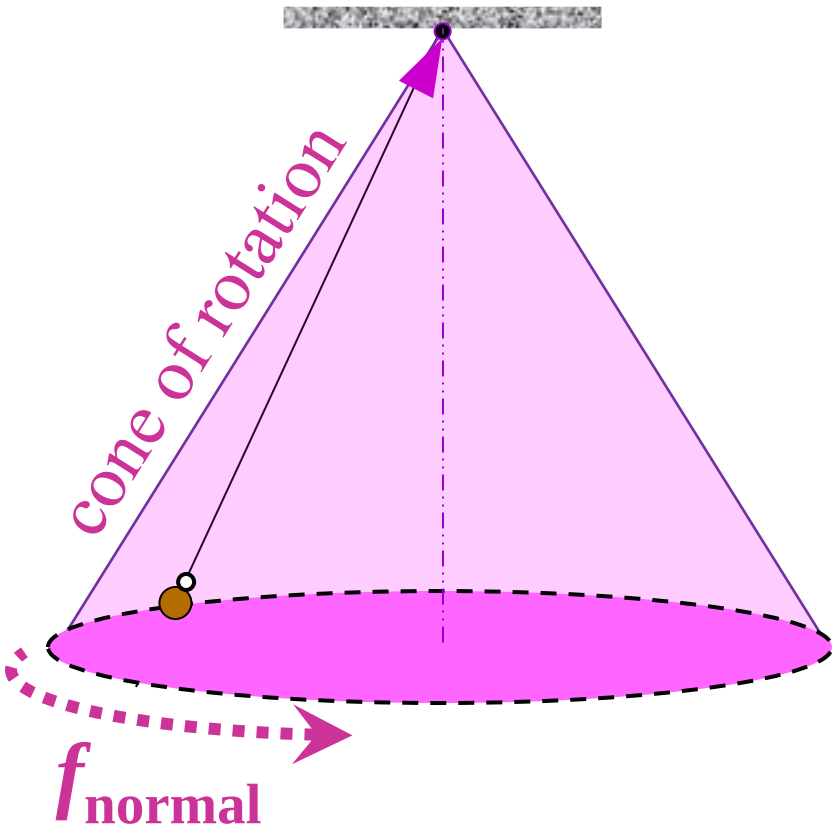




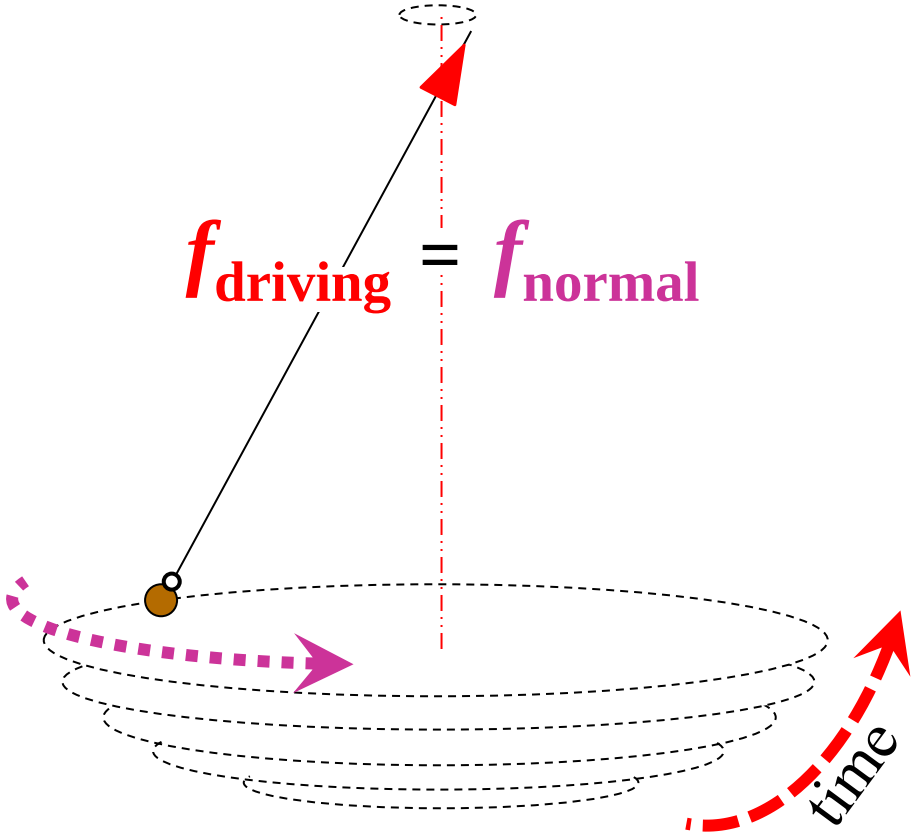


Normal Mode and Resonance of a 2-D Pendulum

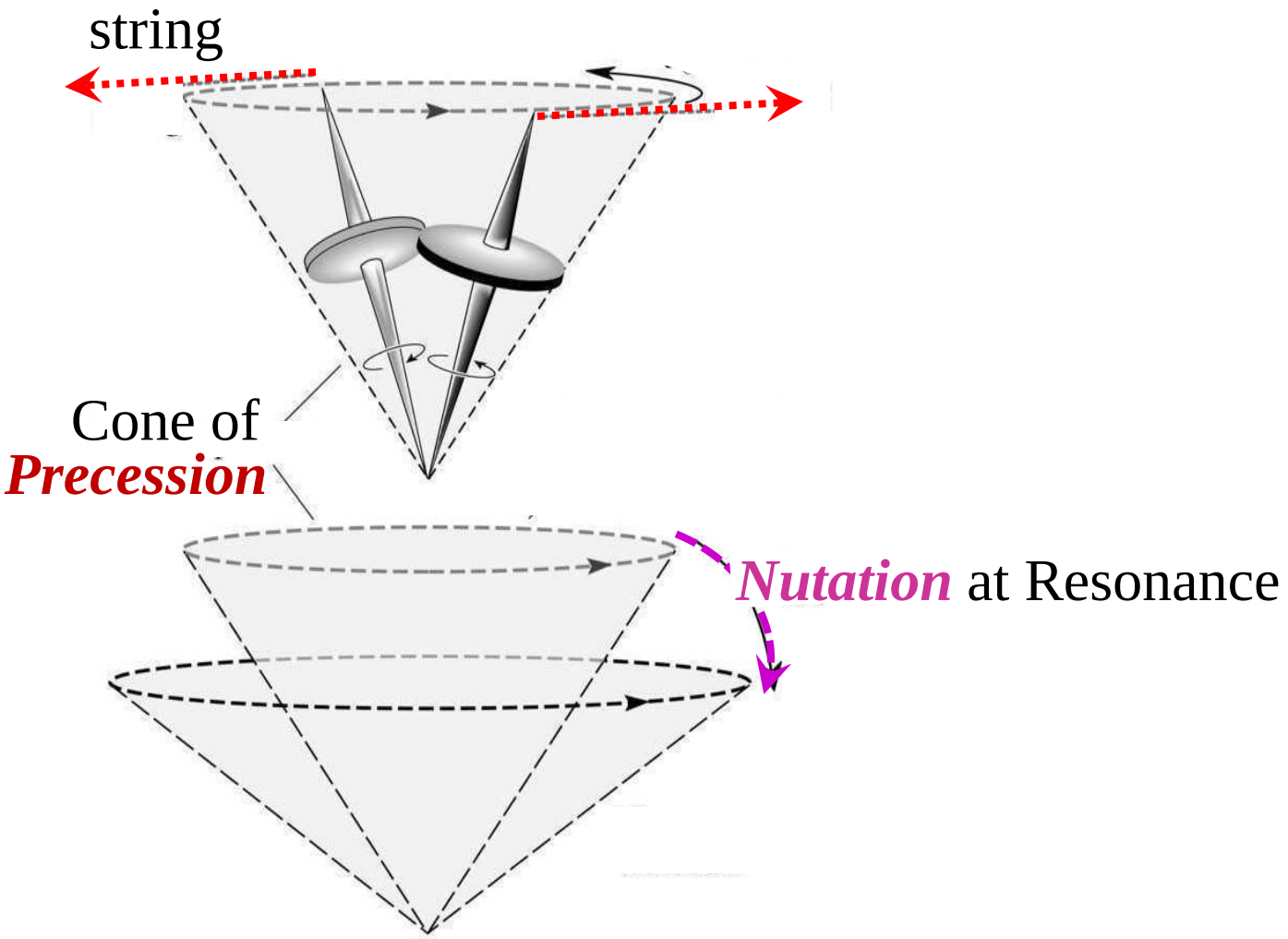
Precession



Nutation

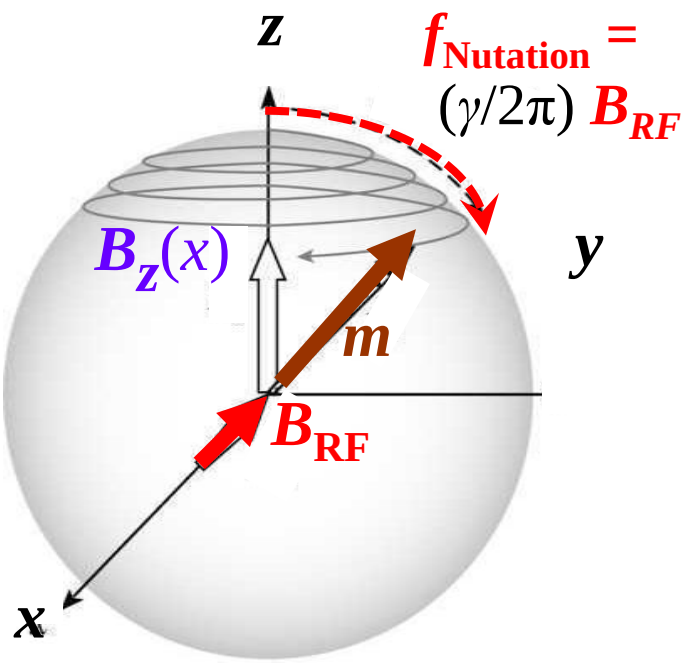


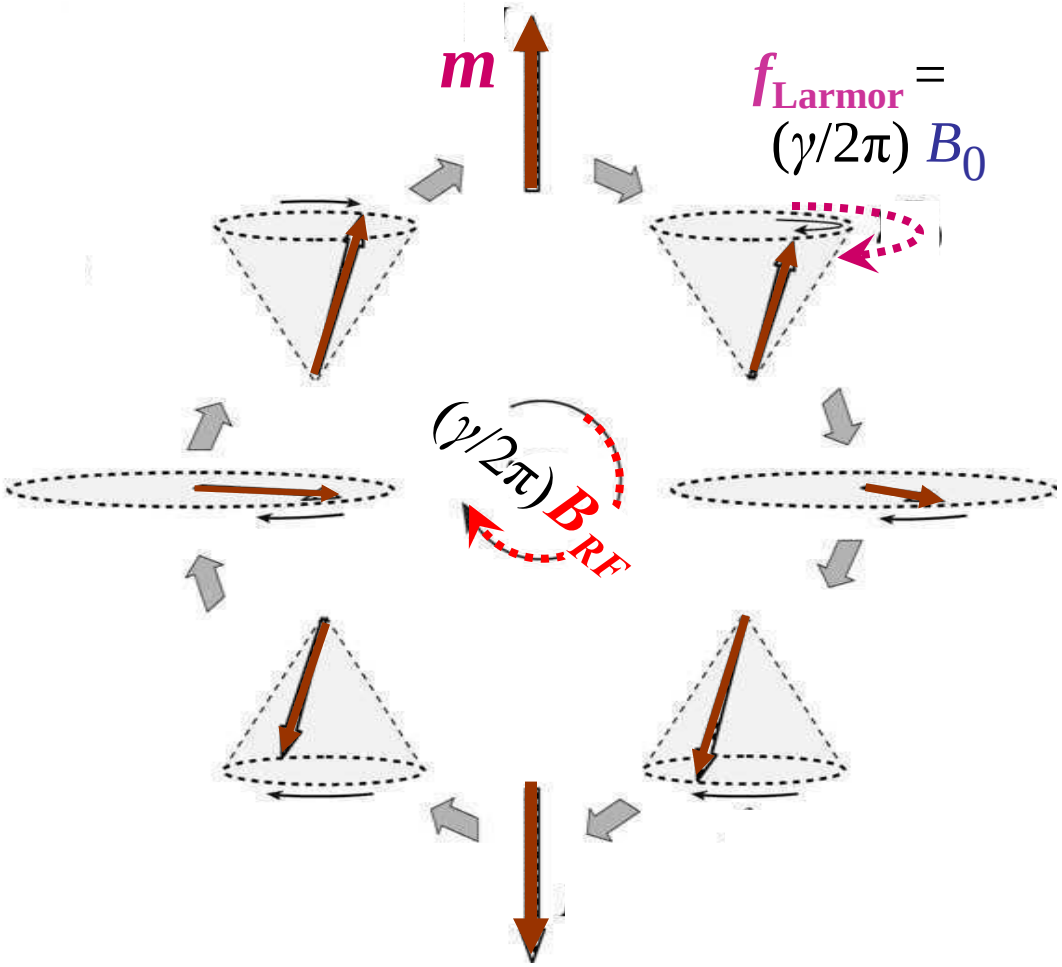
And Now for Something Completely Different: ***Nutation***



Nutation of the Voxel's Magnetization, ***m***(***x***,*t*)

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





precesses at $f_{Lar} = (\gamma/2\pi) B_z(x)$ (always!)

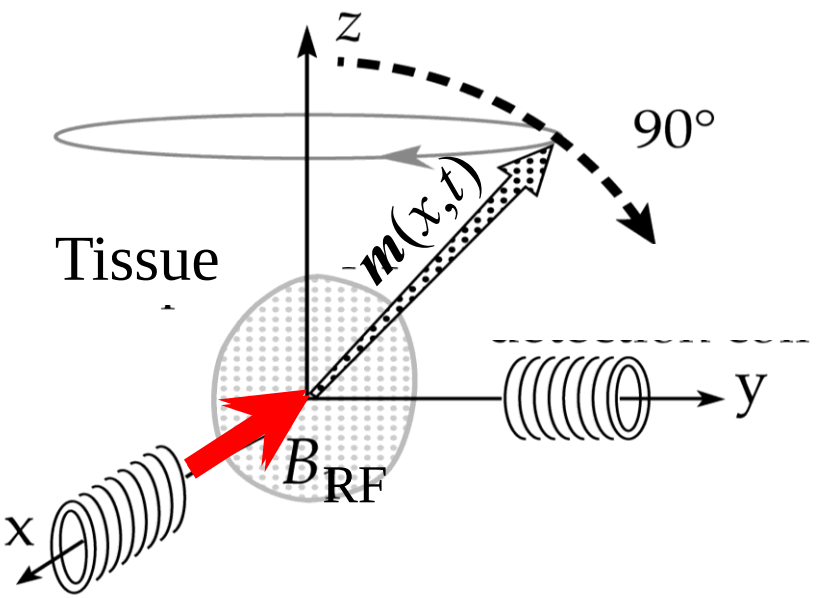
nutates at $f_{nut} = (\gamma/2\pi) B_{RF}$ (only when B_{RF} is on!)

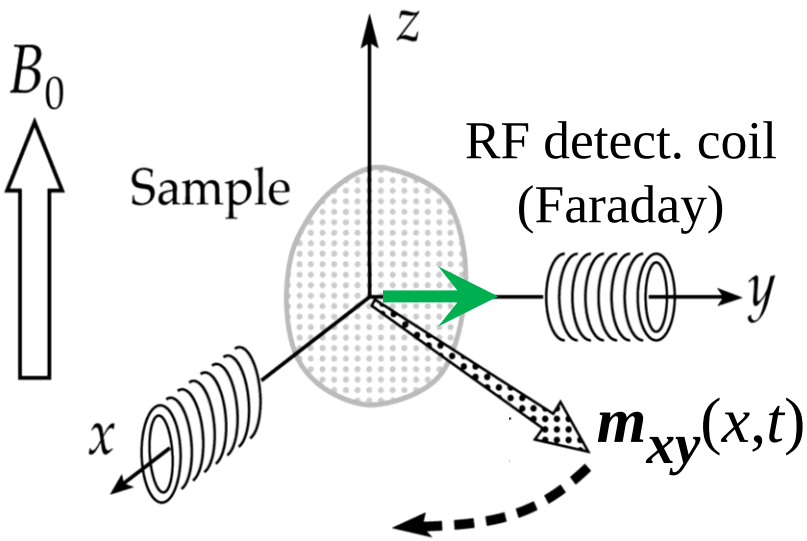
Free Induction Decay (FID) in a voxel, without the decay

e.g., very slow T1 relaxation

$\boldsymbol{m}(x,t)$ for a *Single* Voxel at x and Precessing in the x-y Plane

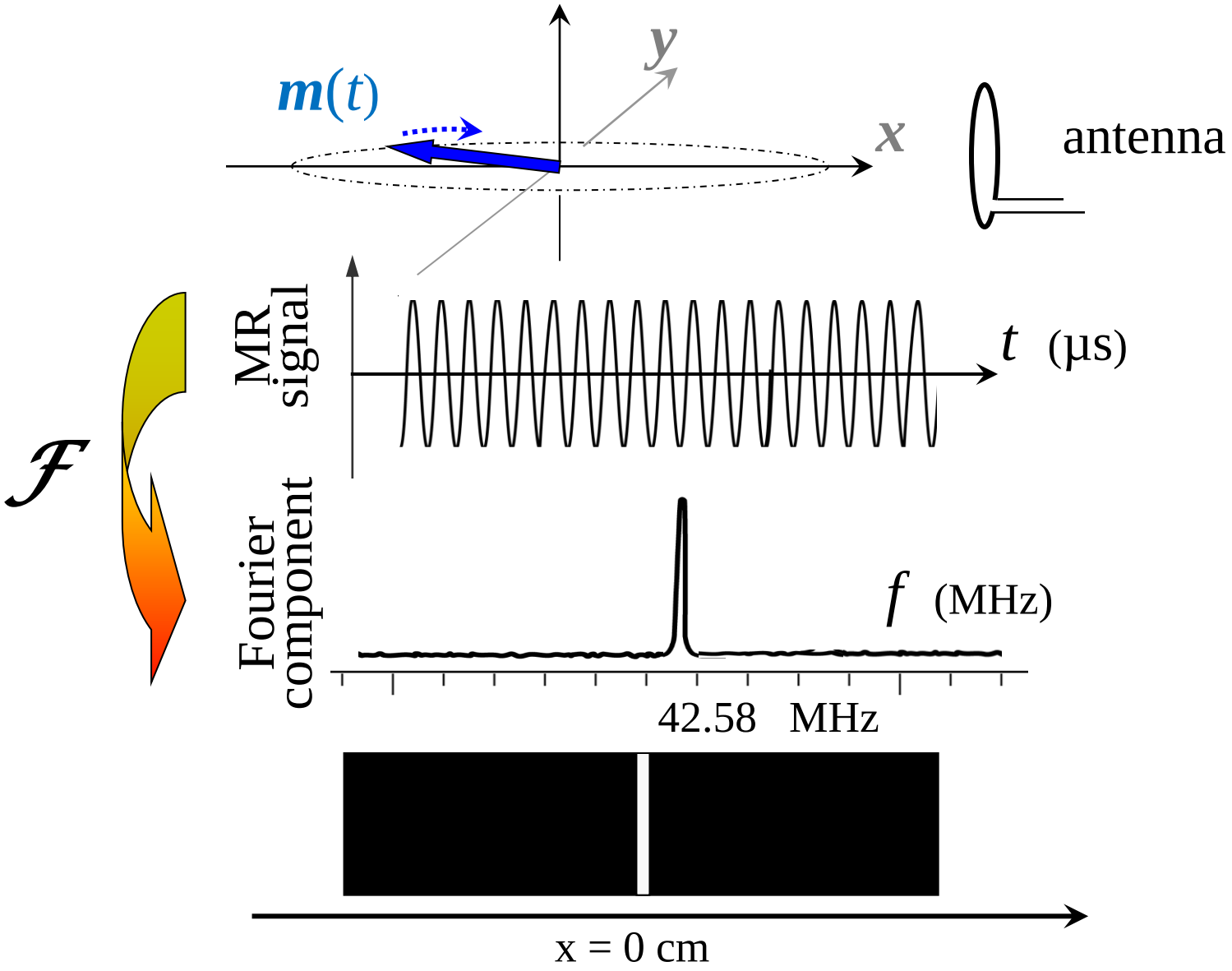
following a 90º pulse





FID Precession, Reception, Fourier Analysis (single voxel)

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



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

!!!

III-6. The rate of magnetization nutation depends upon:

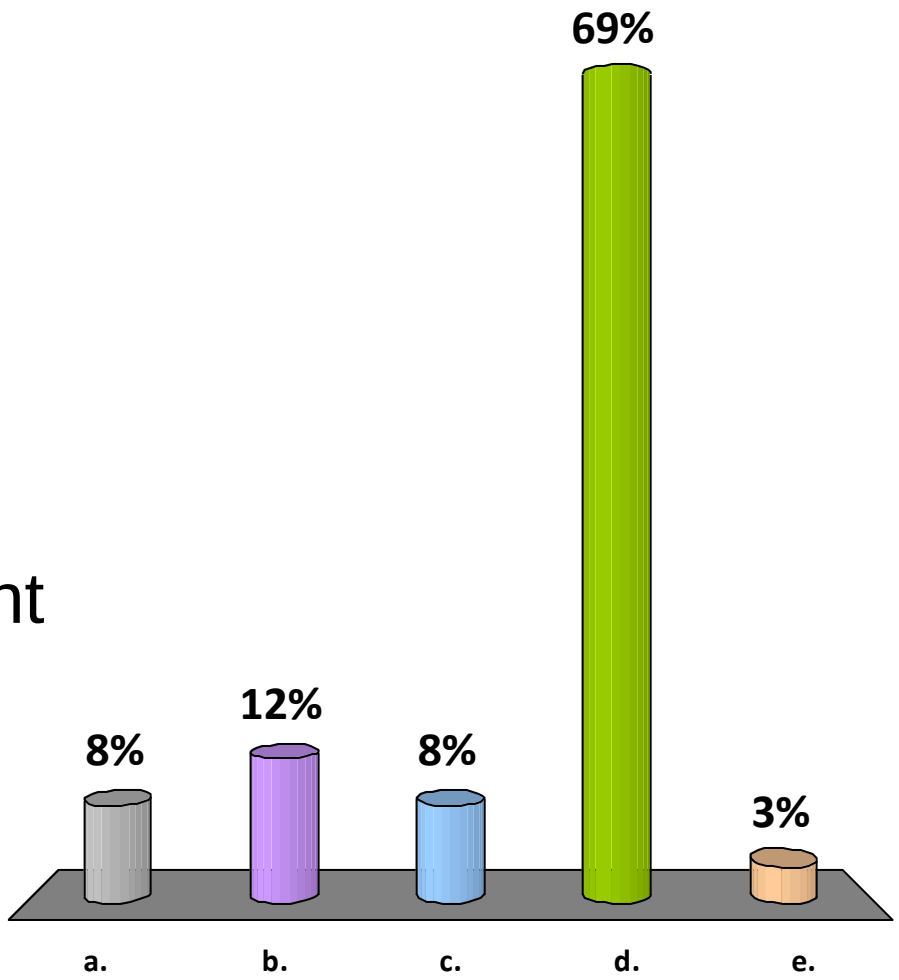
a. the magnitude of the magnetization, |m(t)|

b. the strength of the principal magnetic field, $|B_0|$

c. strength of the *x*-gradient field, $|x \times G_x|$

d. the strength of the RF magnetic field, $|B_{RF}(t)|$

e. T1



SAMs Q:

III-6. The rate of magnetization nutation depends upon:

a) the magnitude of the magnetization, $|\boldsymbol{m}(t)|$

b) the strength of the principal magnetic field, $|\boldsymbol{B}_0|$

c) strength of the $\boldsymbol{x}$-gradient field, $|\boldsymbol{x} \times G_{\boldsymbol{x}}|$

d) the strength of the RF magnetic field, $|\boldsymbol{B}_{\mathrm{RF}}(t)|$

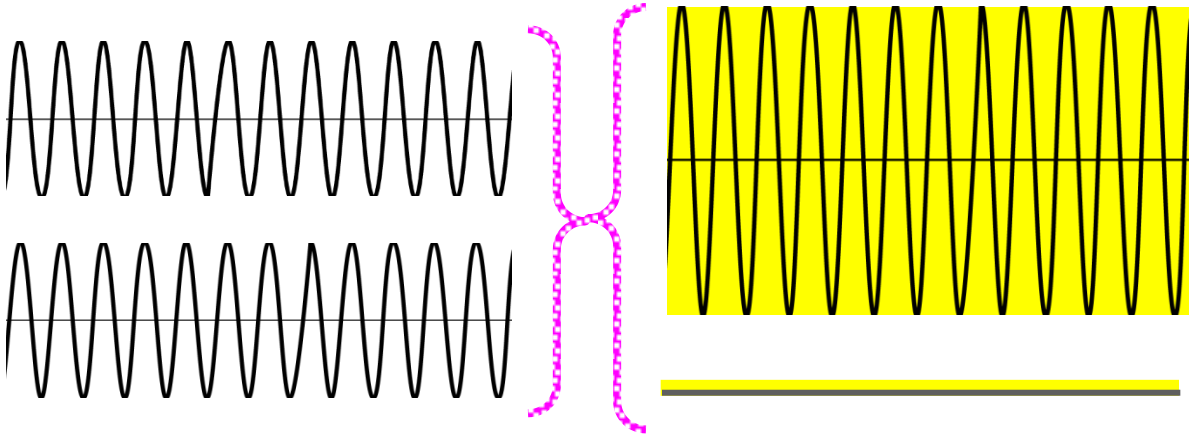
e) T1

Answer: (d).

Ref: "Medical Imaging", A.B. Wolbarst *et al.*, Wiley-Blackwell (2013), p. 328.

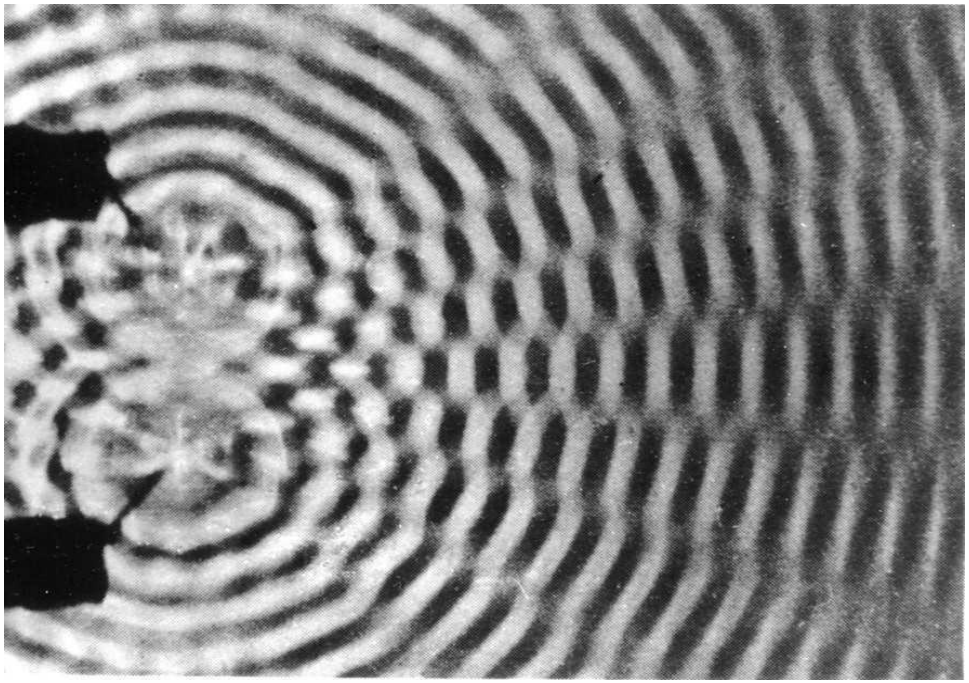
Untangling the FID RF Signal from Two Voxels: temporal Fourier series approach

Wave Interference in Time or Space

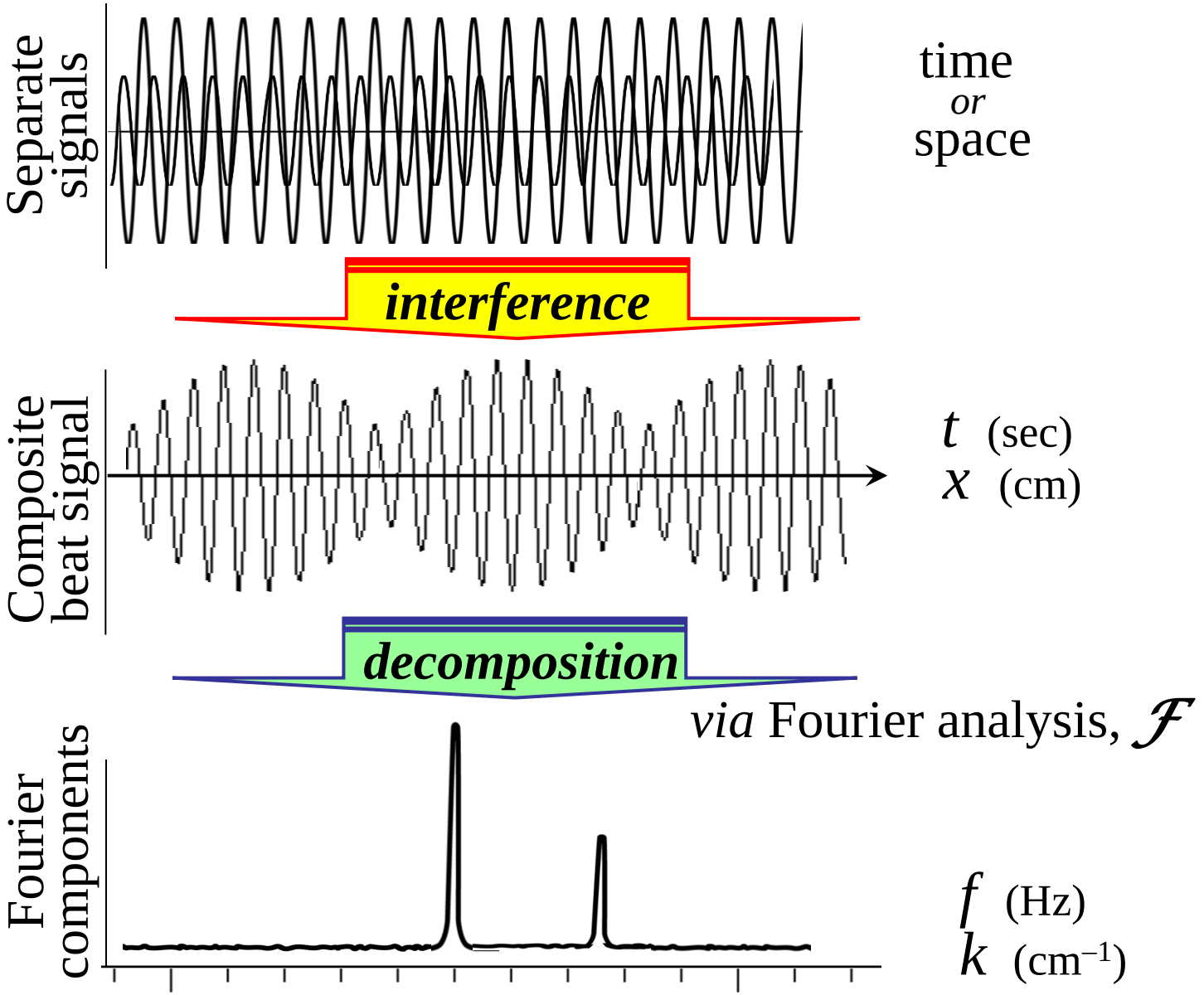


positive
(in-phase)
interference

negative
interference



Wave Interference and Spectral Analysis

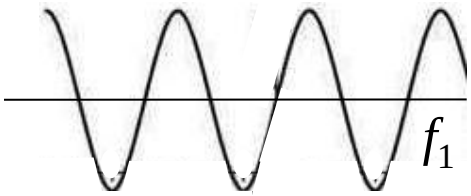


Fourier Decomposition of Periodic Temporal Signal

$S(t) \sim \frac{1}{2} + (2/\pi)\{\sin(2\pi f_1 t) + \frac{1}{3}\sin(6\pi f_1 t) + 1/5 \sin(10\pi f_1 t) + \ldots\}$

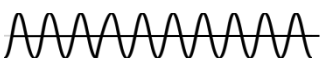
fundamental: f_1 Hz (cycles/sec)

orthonormal basis vectors

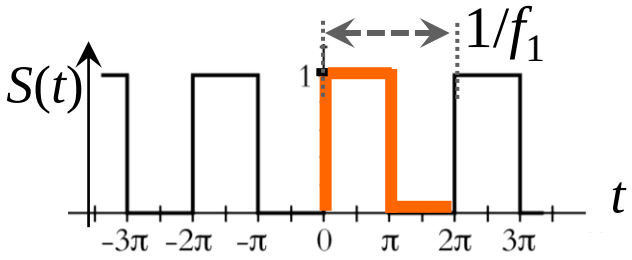


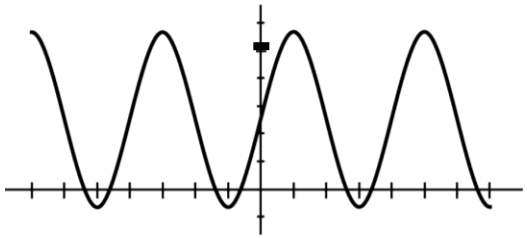
fundamental

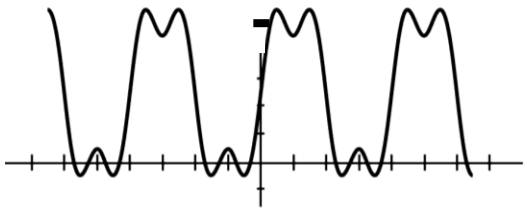
3rd harmonic

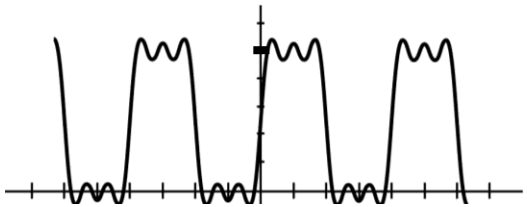


5th

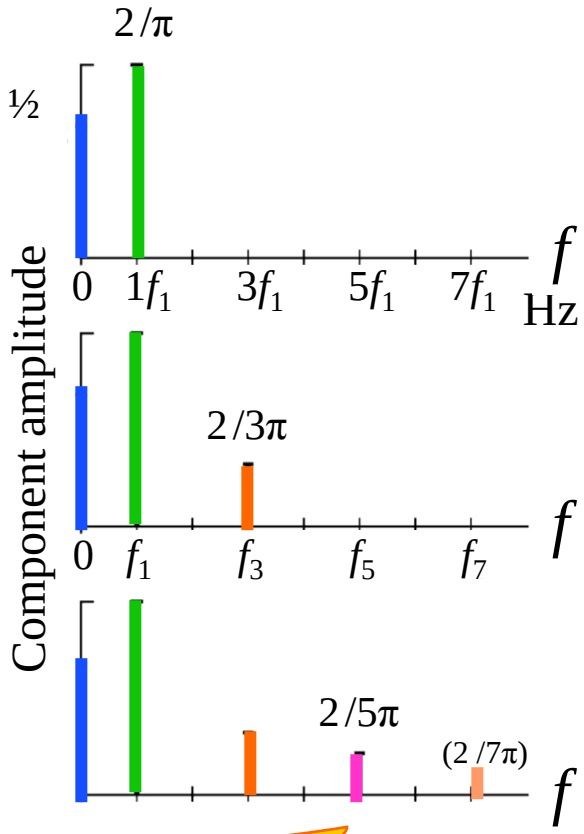








spectrum

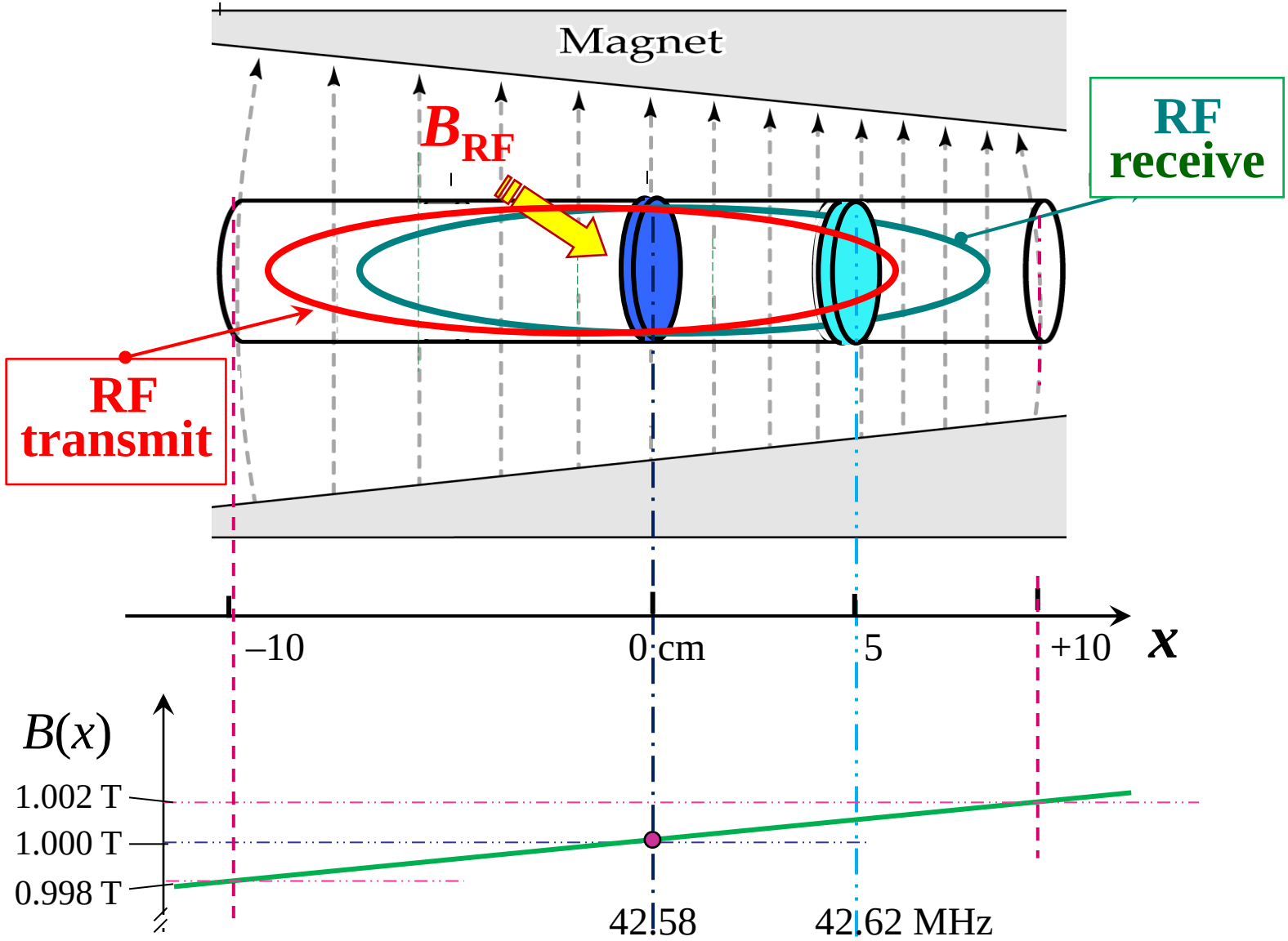


$\mathcal{F}$

FID imaging of the two-voxel 1D patient

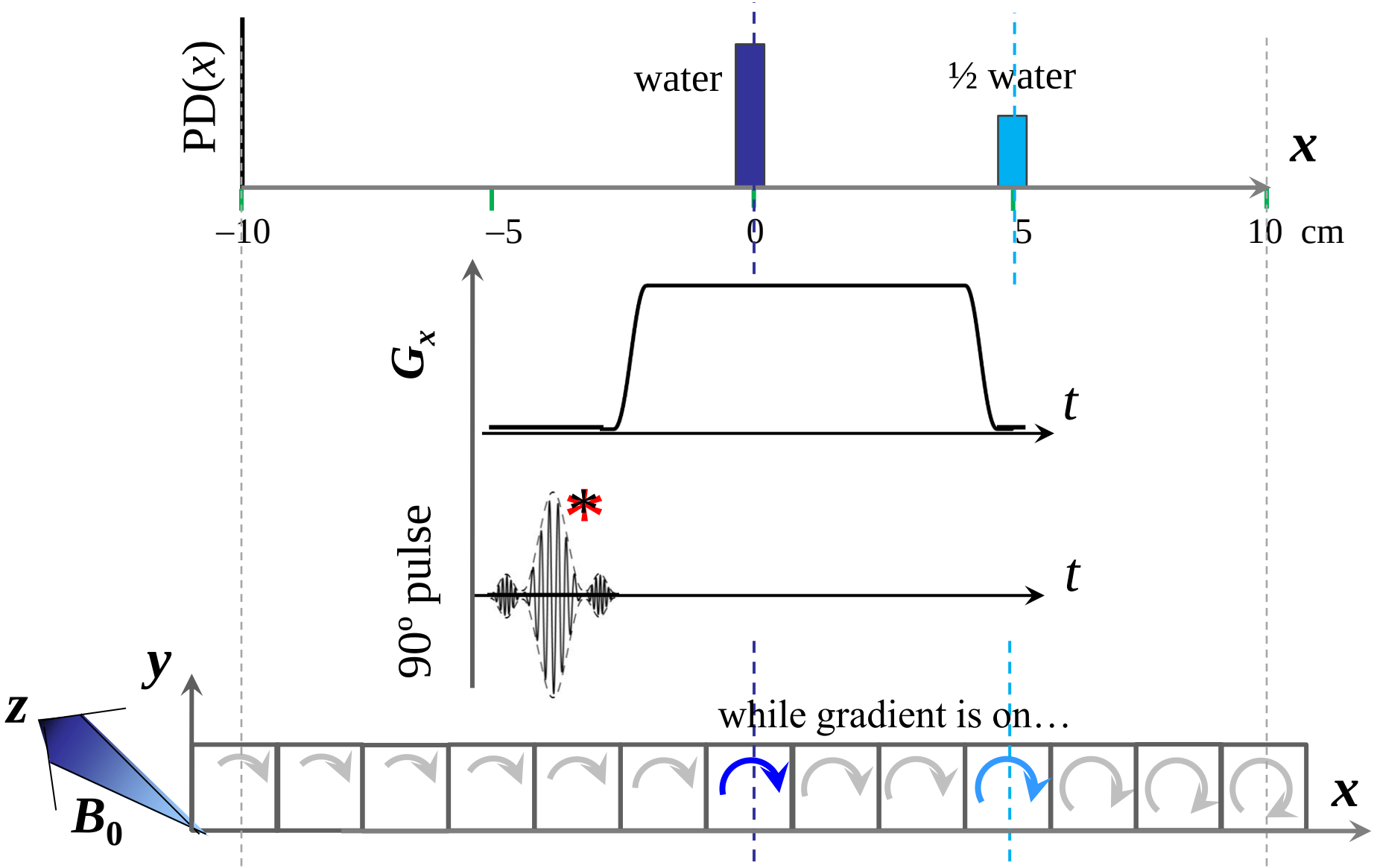
$f_{\text{Larmor}}(x) = (\gamma/2\pi)\,(x \times G_x)$ (in rotating frame)

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



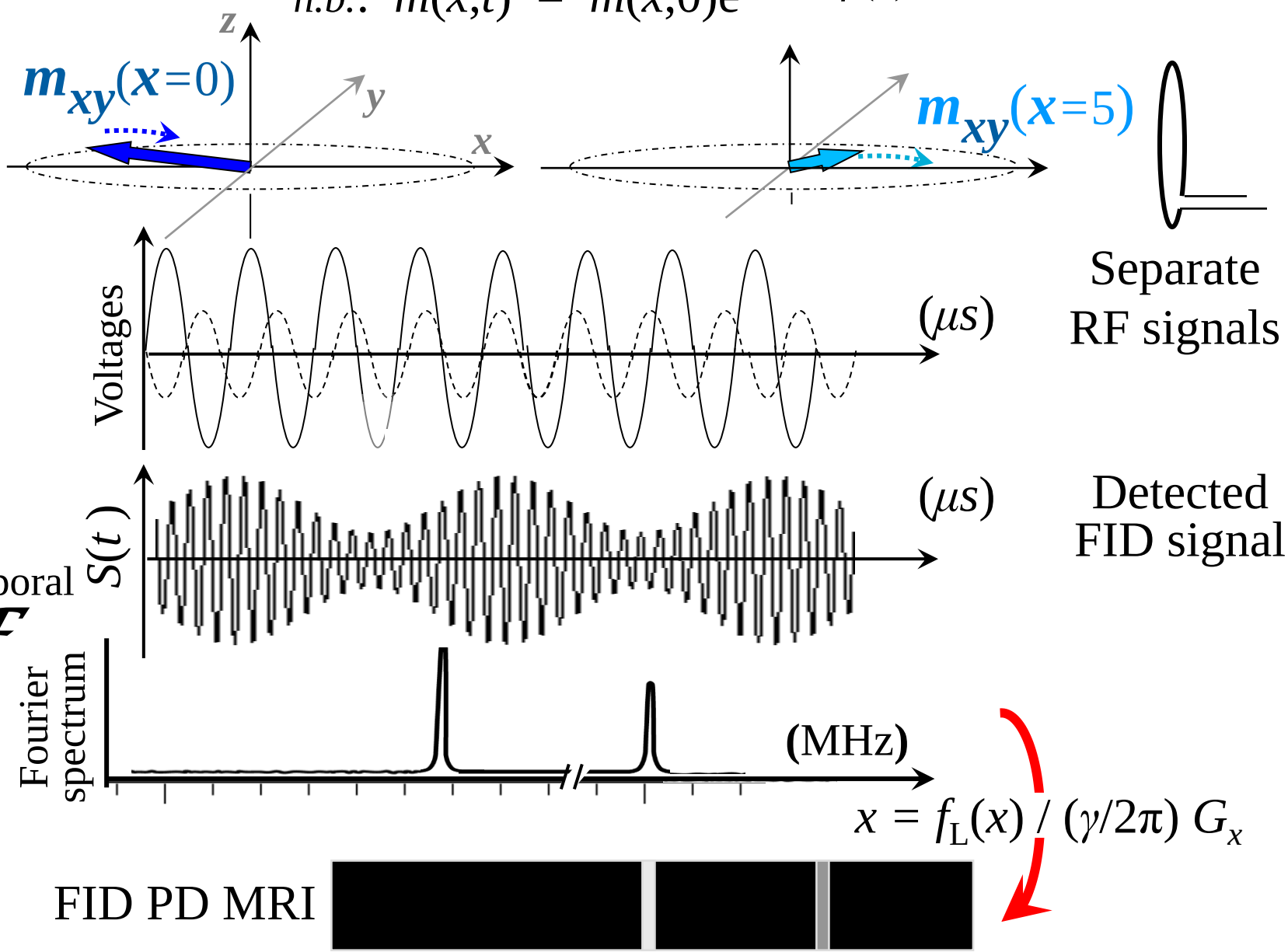
FID Following *Wide-Bandwidth** 90° RF Pulse

covers $f_{\text{Larmor}}(x)$ for all N voxels while $\boldsymbol{G_x}$ is being applied!



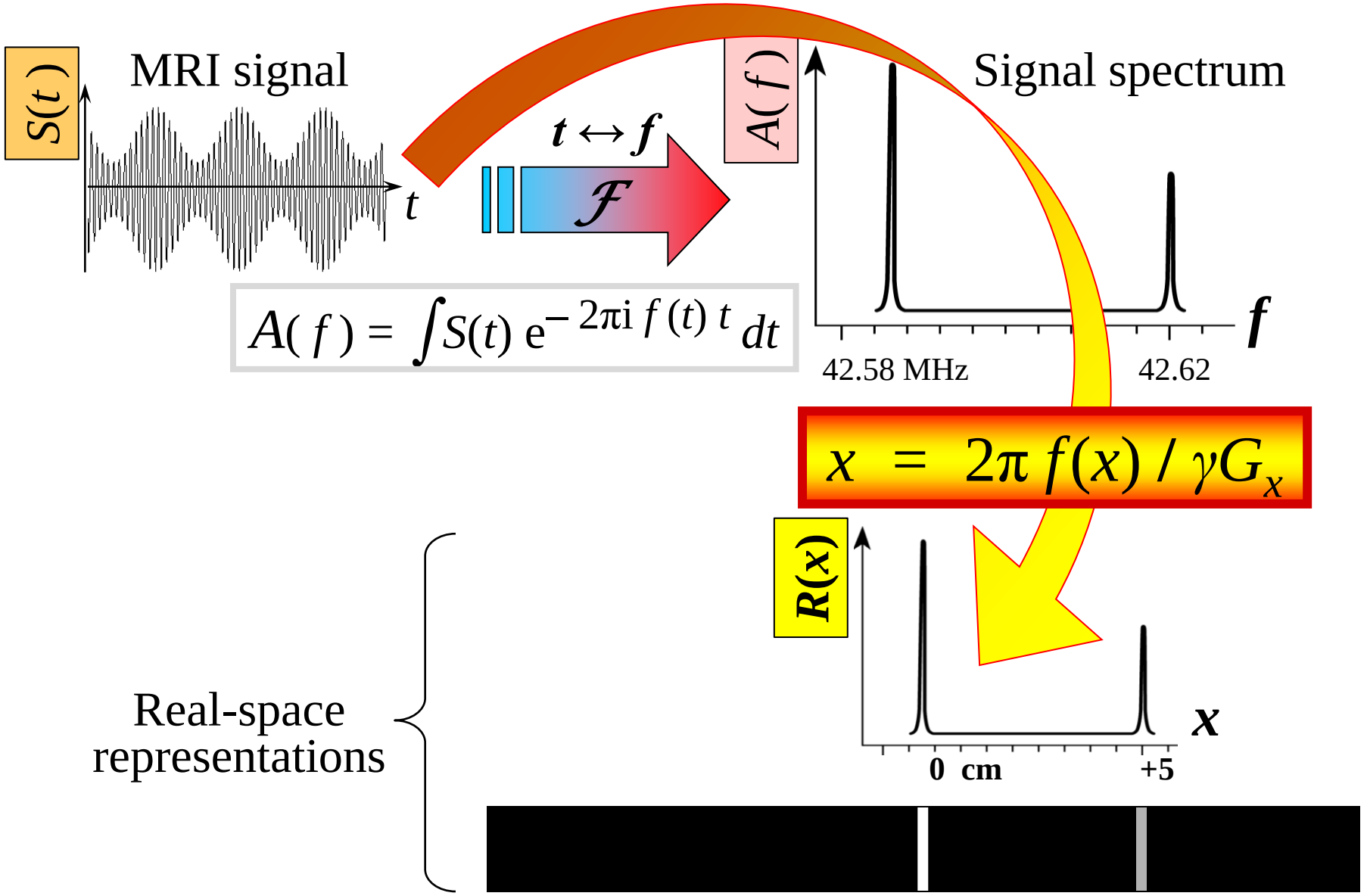
FID: After Wide-Band RF 90º Pulse Centered on $f_{\text{Larmor}}(B_0)$

$$n.b.:\ m(x,t) = m(x,0)e^{-2\pi i f(x) t}$$



To Summarize What We Have Done So Far with FID

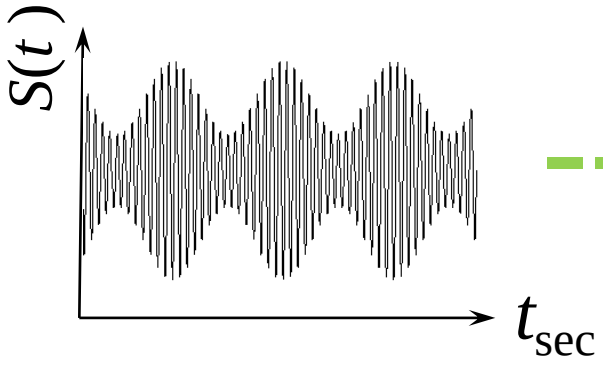
follow temporal $\mathcal{F}$ with isomorphism of $A(f)$ to real-space, $R(x)$



FID Imaging *via* ***k***-Space; *Spatial* FT

Again, What We Have Done So Far:

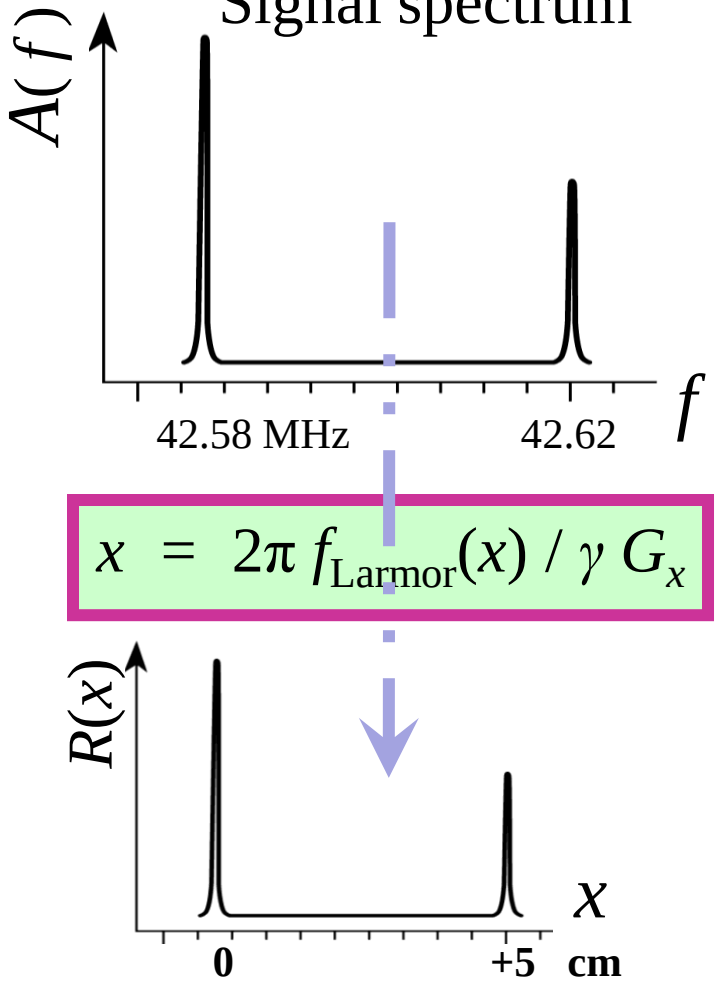
Detected FID signal



Temporal:
$t \leftrightarrow f$

$\mathcal{F}$

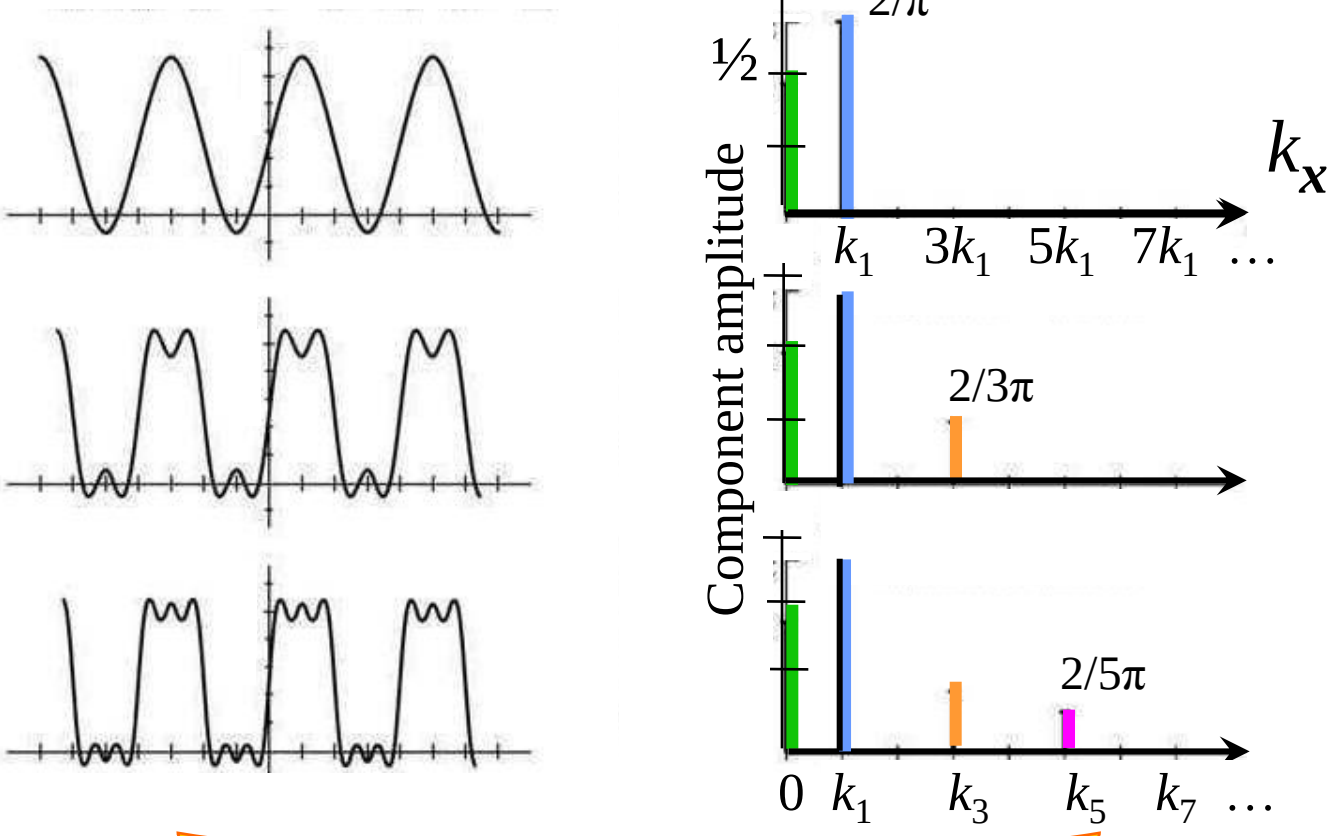
Signal spectrum



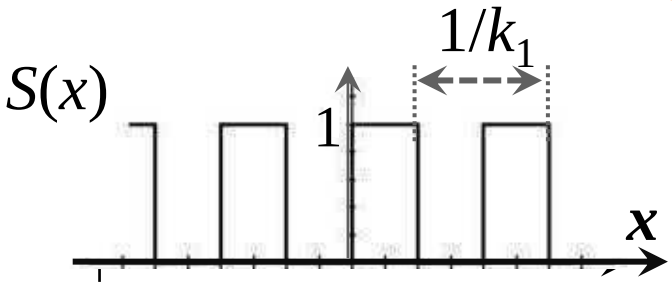
Real-space representation

Now Consider 1D ***Spatial*** Fourier Analysis

fundamental: k_1 (cycles/mm)

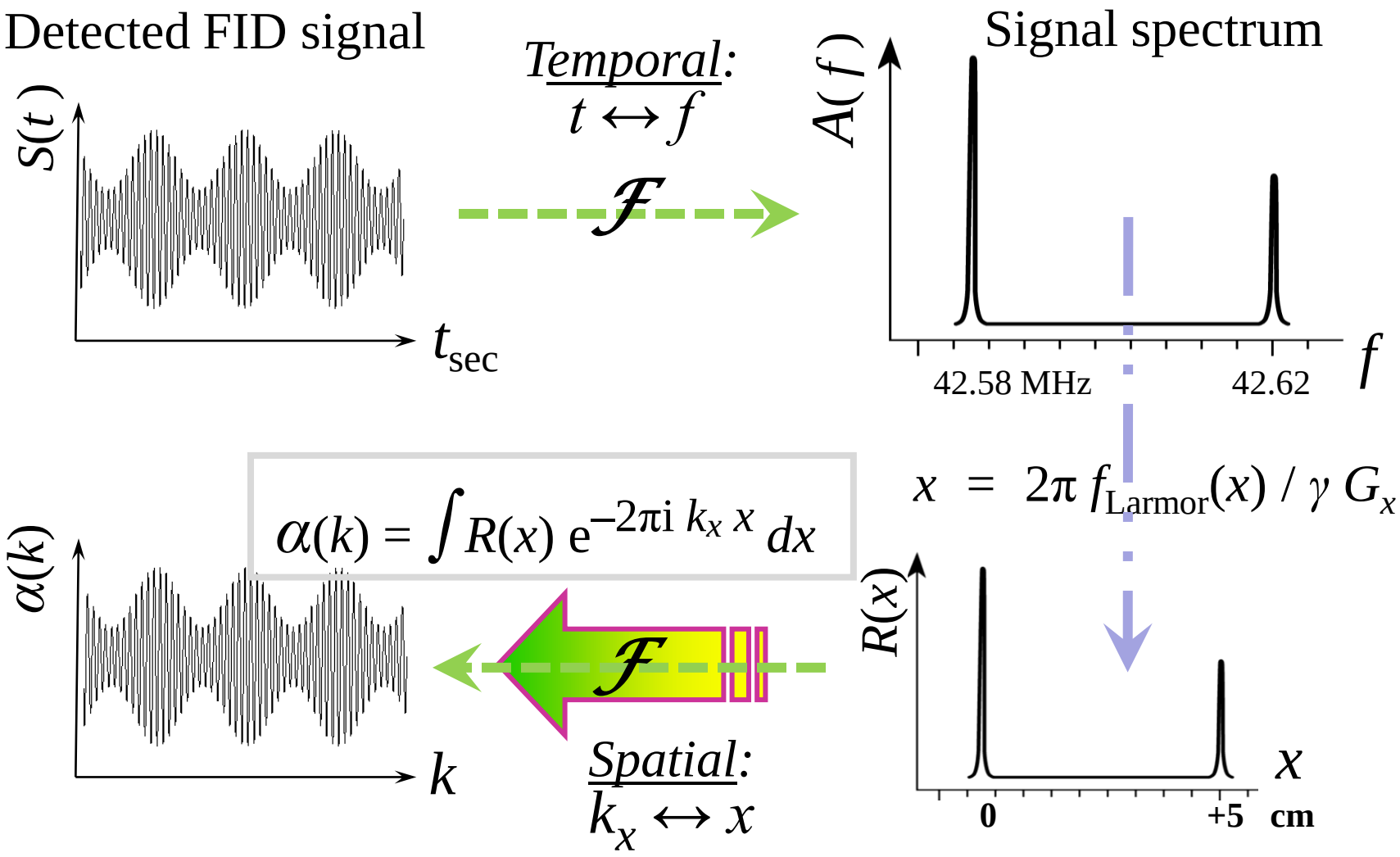


$\mathcal{F}$

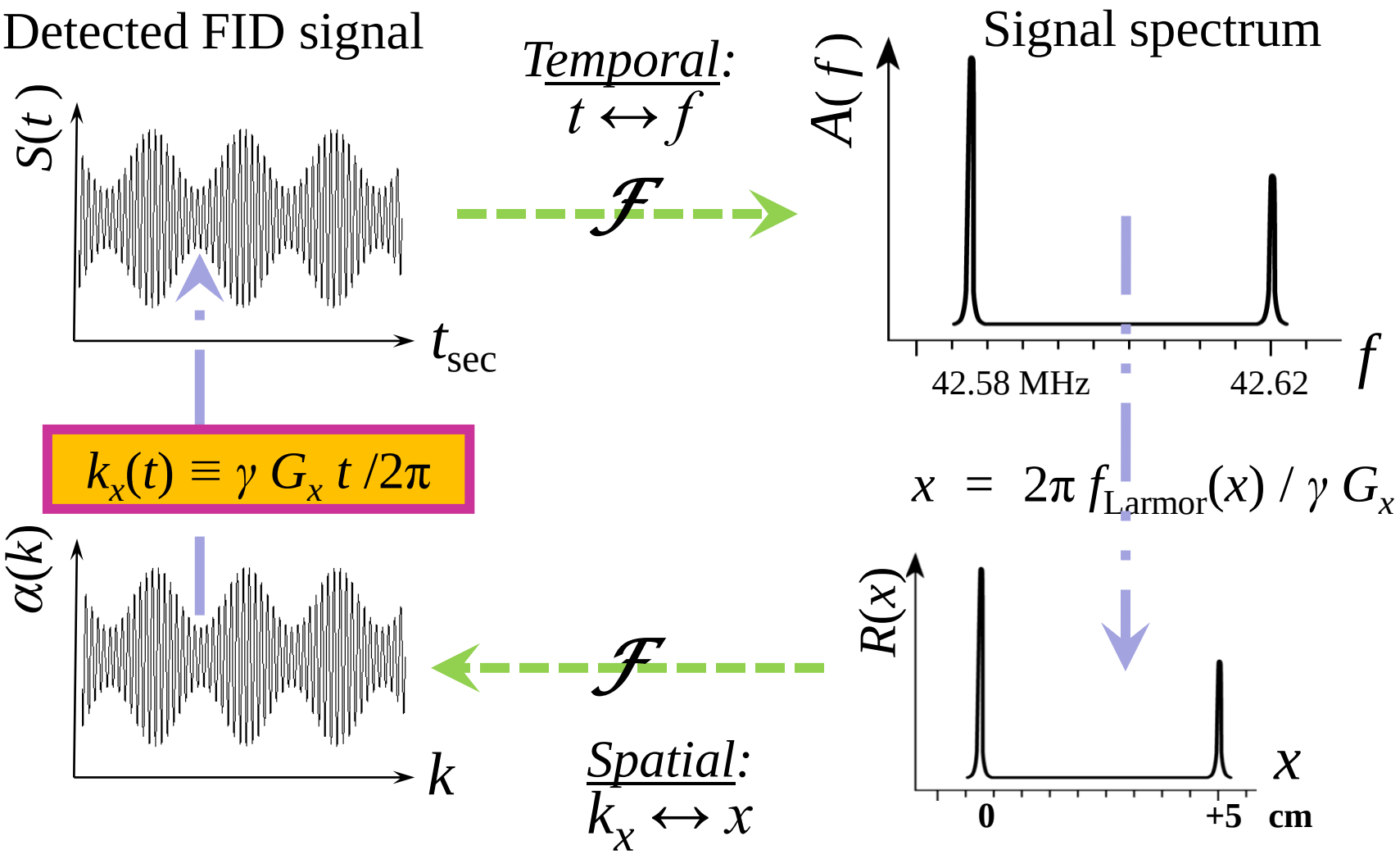


$$S(x) \sim \tfrac{1}{2} + (2/\pi)\,[\sin(2\pi k_1 x) + \tfrac{1}{3}\sin(6\pi k_1 x) + 1/5\sin(10\pi k_1 x) + \ldots]$$

Experiment: Continue on with a FT from x- to k_x-Space

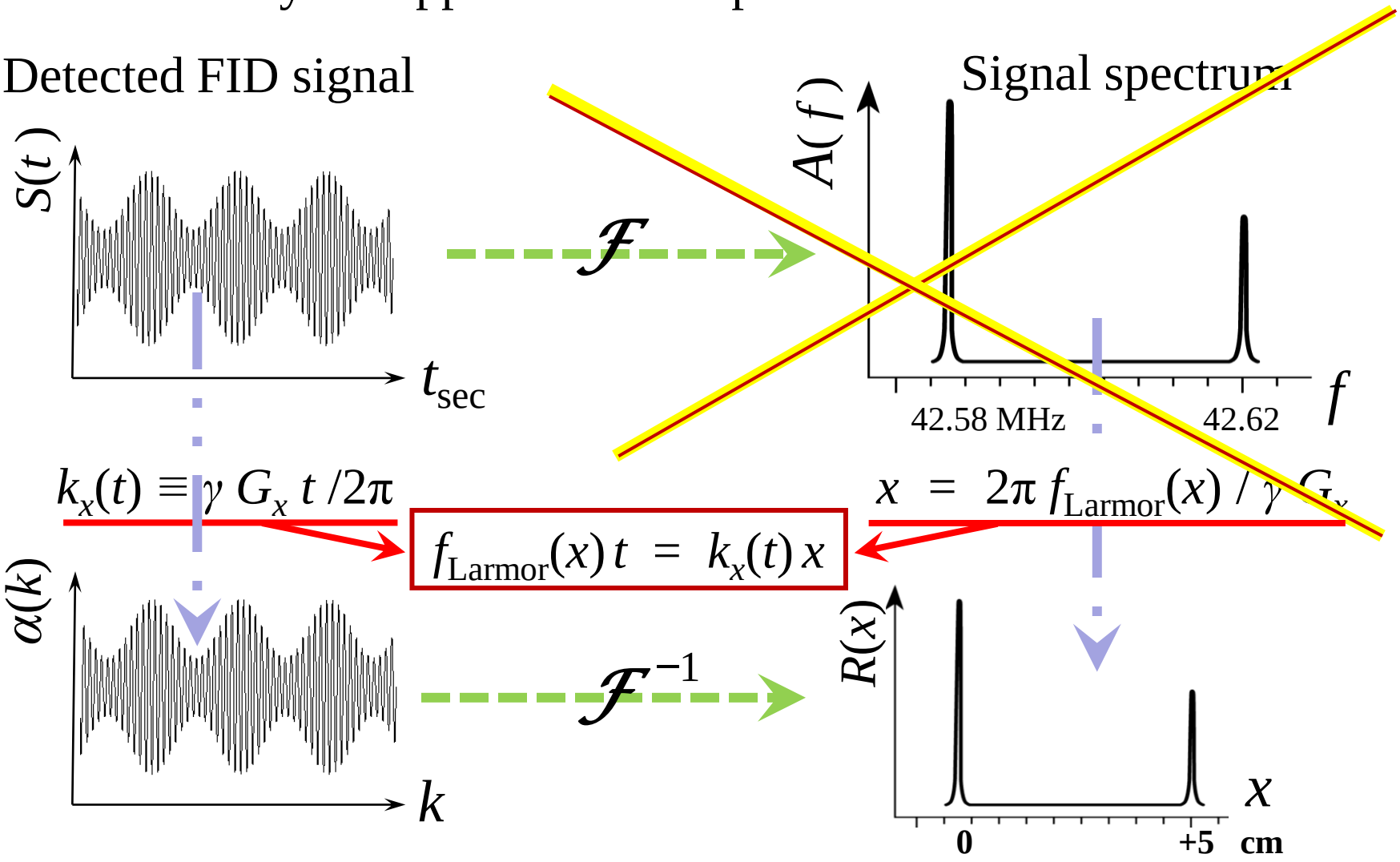


And Complete the Loop Linking t-, f-, x- and k_x-Spaces



So, We Can Get from $S(t)$ to $R(x)$ in Either of Two Ways!

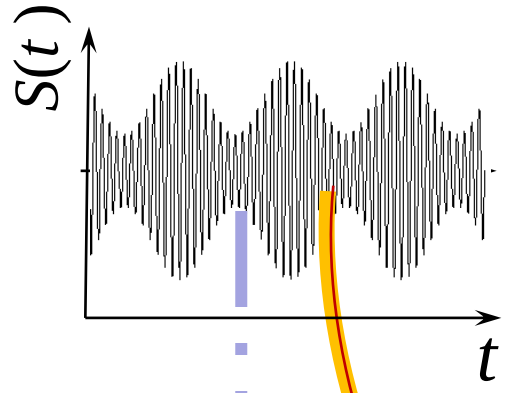
but only the approach *via* ***k***-space works in 2D and 3D



k-Space Procedure for FID in 1D Phantom

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

Detected FID signal



$$m(x,t) = m(x,0)\mathrm{e}^{-2\pi \mathrm{i}\, f(x)\, t} = m(x,0)\, \mathrm{e}^{-2\pi i\, k_x(t)\, x}$$

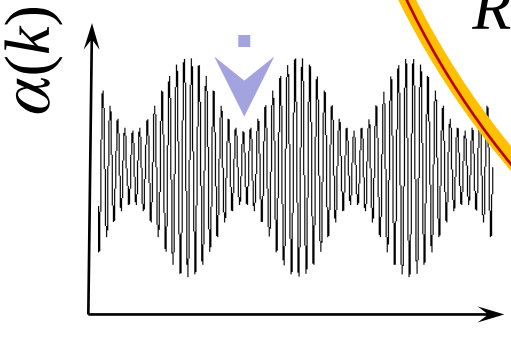
$$S(t) = S(k(t)) \sim \int m(x)\, \mathrm{e}^{-2\pi i\, k_x(t) x}\, dx\ .$$

Therefore, taking the inverse FT, $\mathcal{F}^{-1}$:

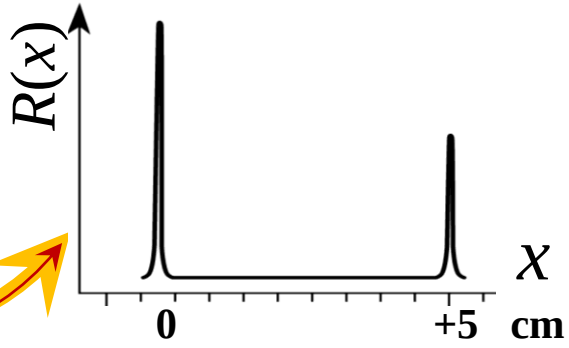
$$m(\boldsymbol{x}) = \int S(t)\, \mathrm{e}^{+2\pi \mathrm{i}\, \boldsymbol{k}(t)\cdot\boldsymbol{x}}\ d^3k$$

$$k_x(t) \equiv \gamma\, G_x\, t\, /2\pi$$

$$R(x) = \int \alpha(k)\, \mathrm{e}^{+2\pi \mathrm{i}\, kx}\, \mathrm{d}k$$



$\mathcal{F}^{-1}$

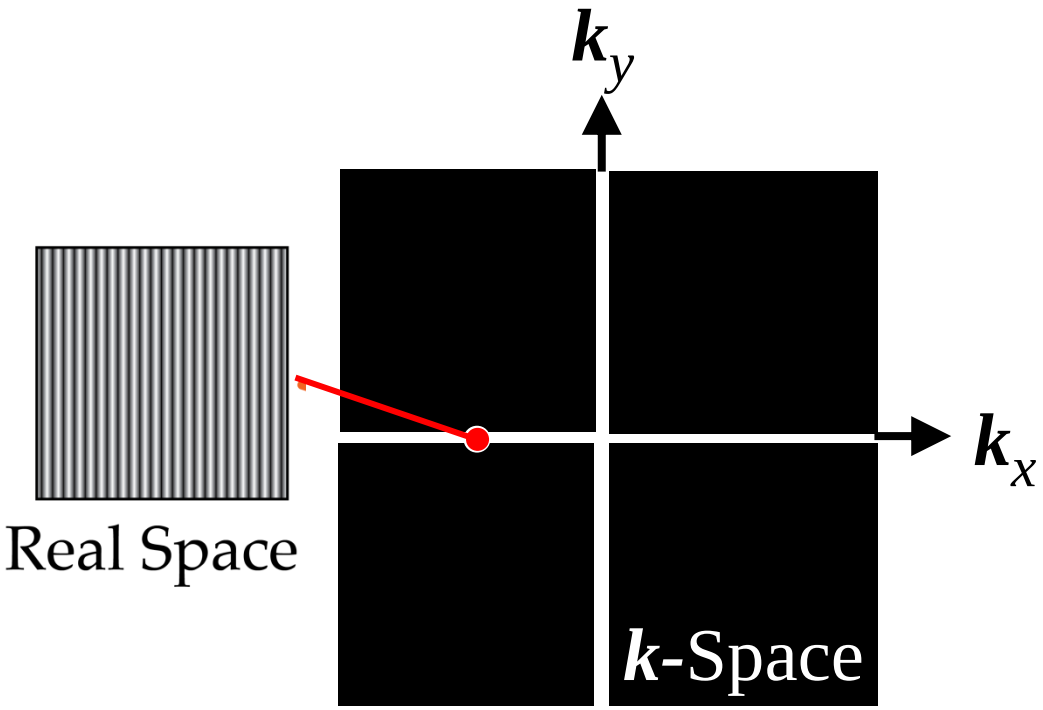


k-space representation

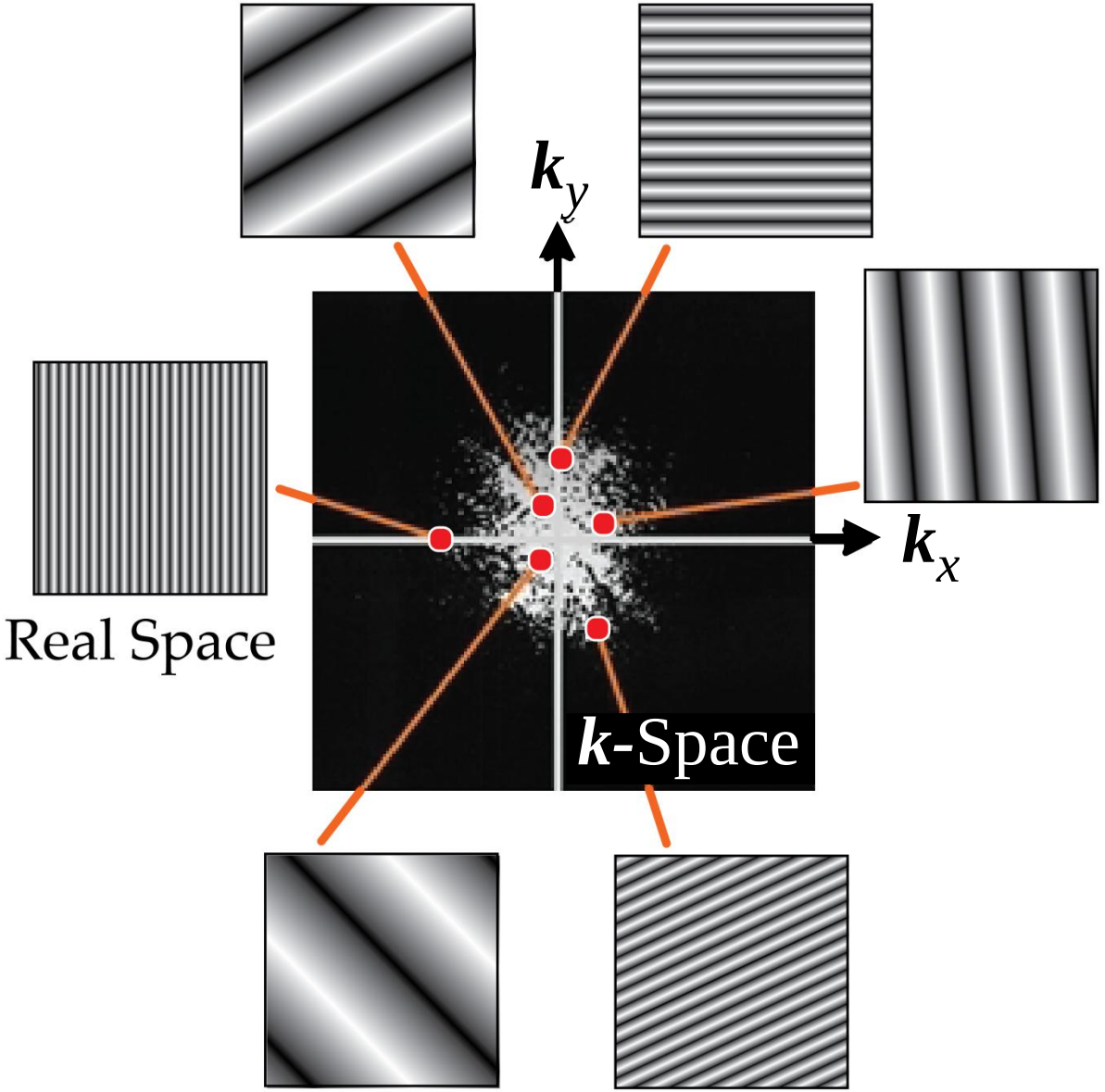
Real-space representation

Spatial Waves, and Points in ***k***-Space

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

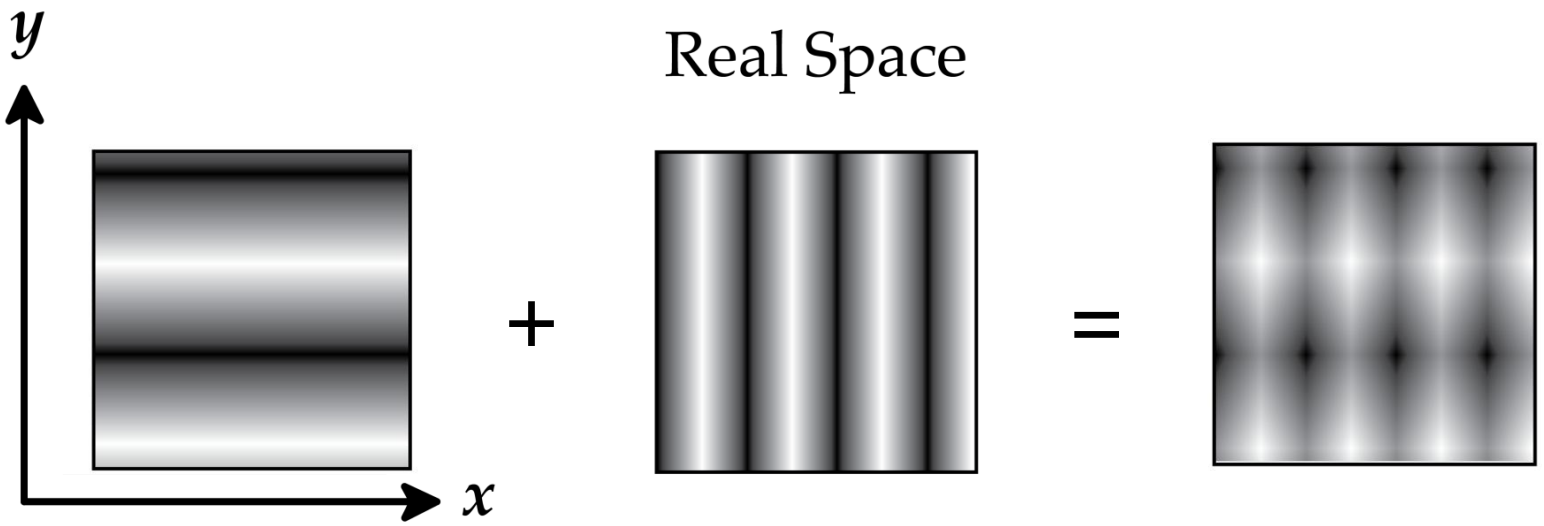


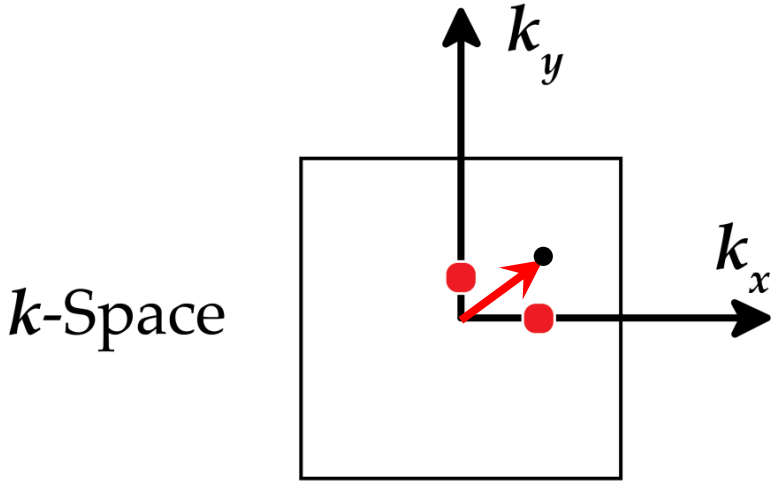
Points in *k*-Space and Their Waves in Real Space



Composite Patterns in *k*-Space and Real Space

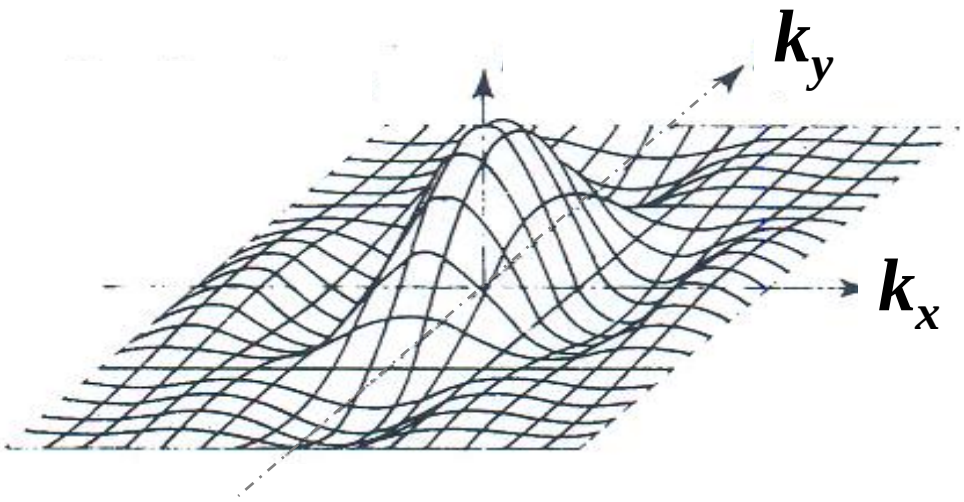
linearity



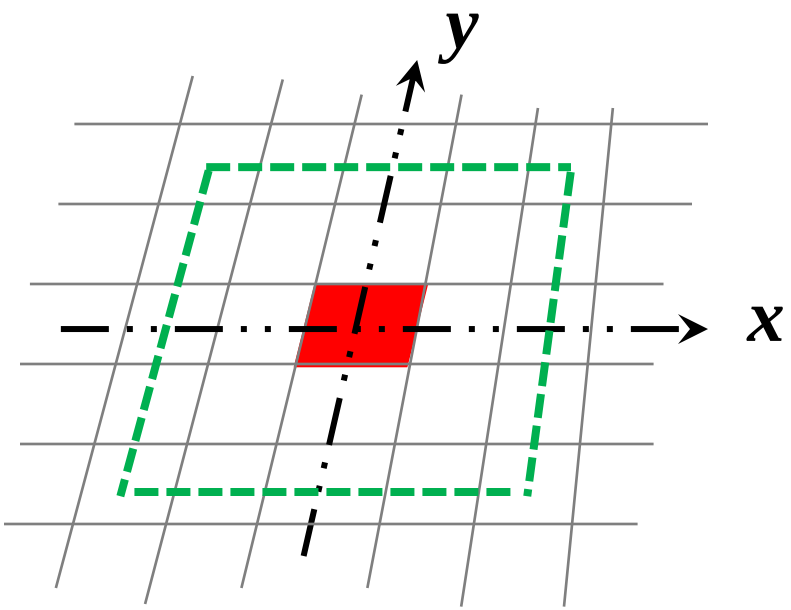


Fourier Transform of *k*-Space Pattern to Real Space

***k*-Space**

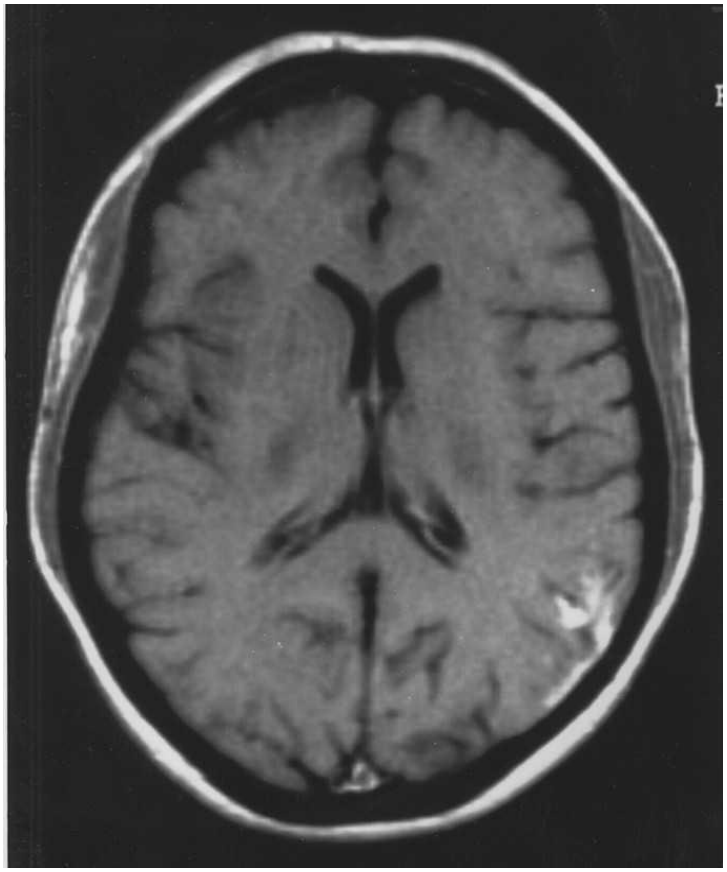


Real Space



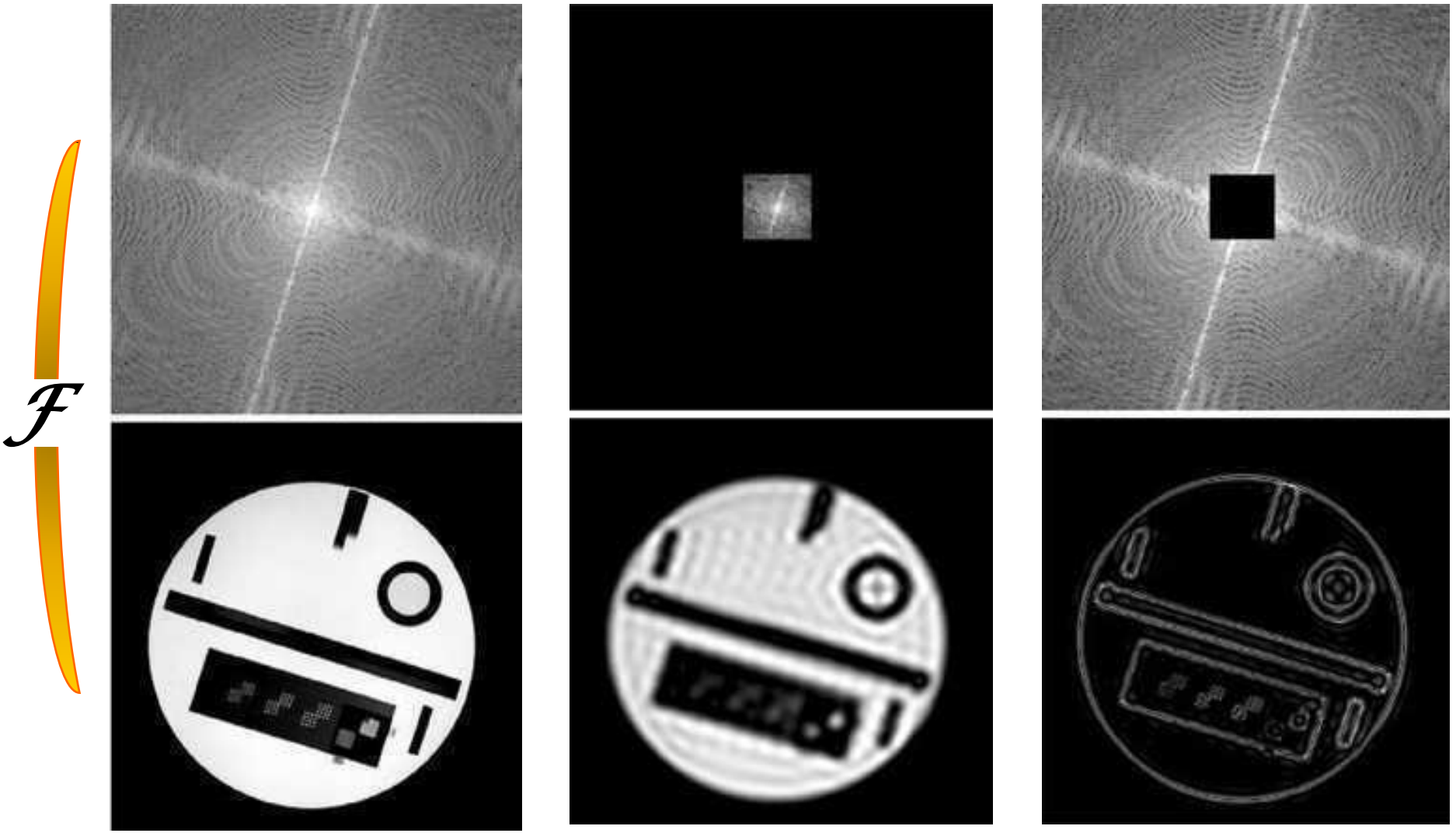
$\mathcal{F}$

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



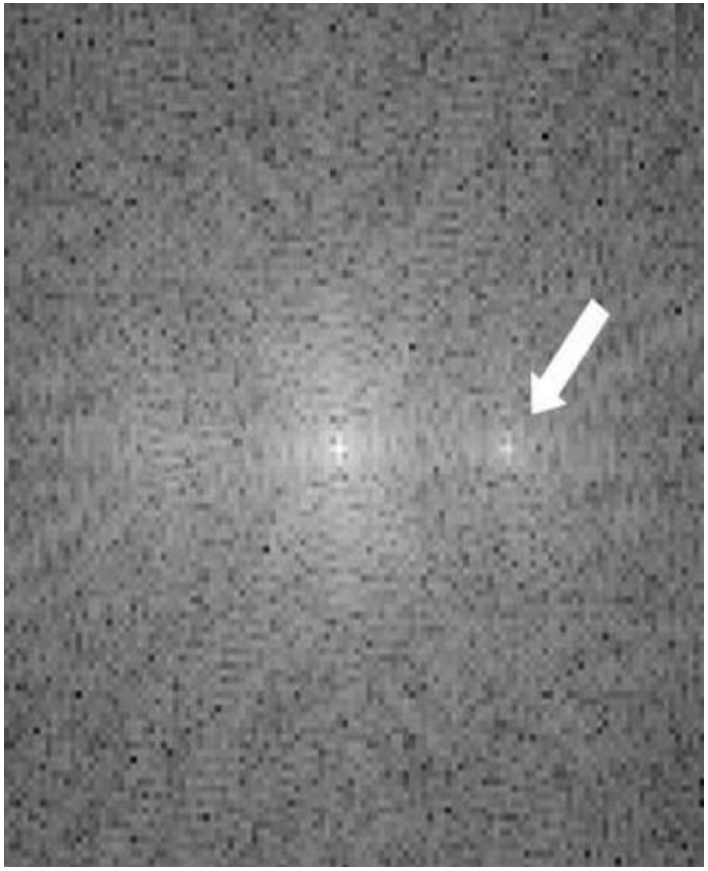
$\mathcal{F}$

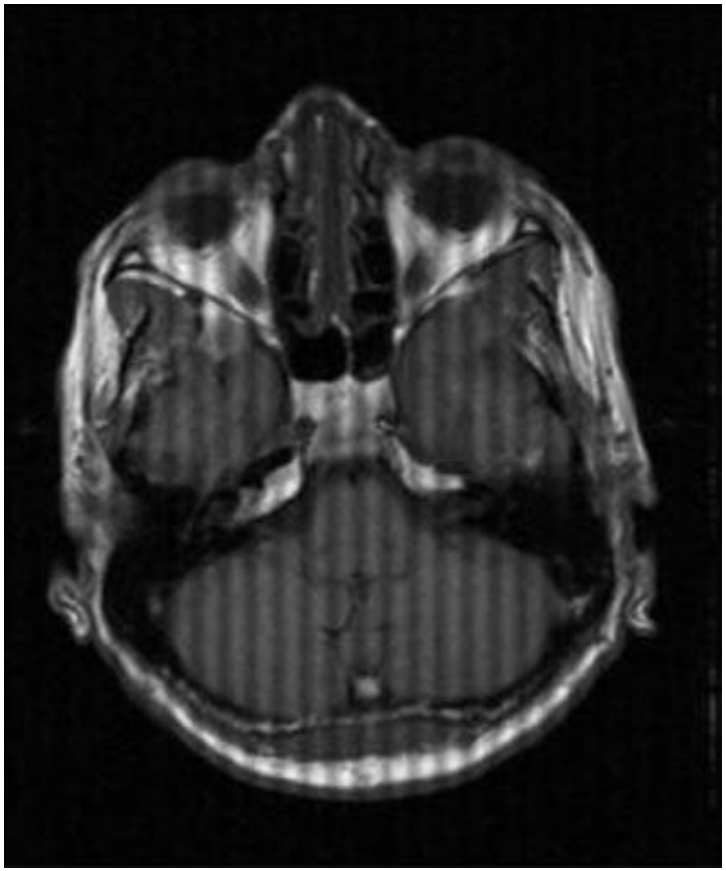
Fourier Transform from Parts of *k*-Space to Real Space



Herringbone Artifact

noise spike during data acquisition

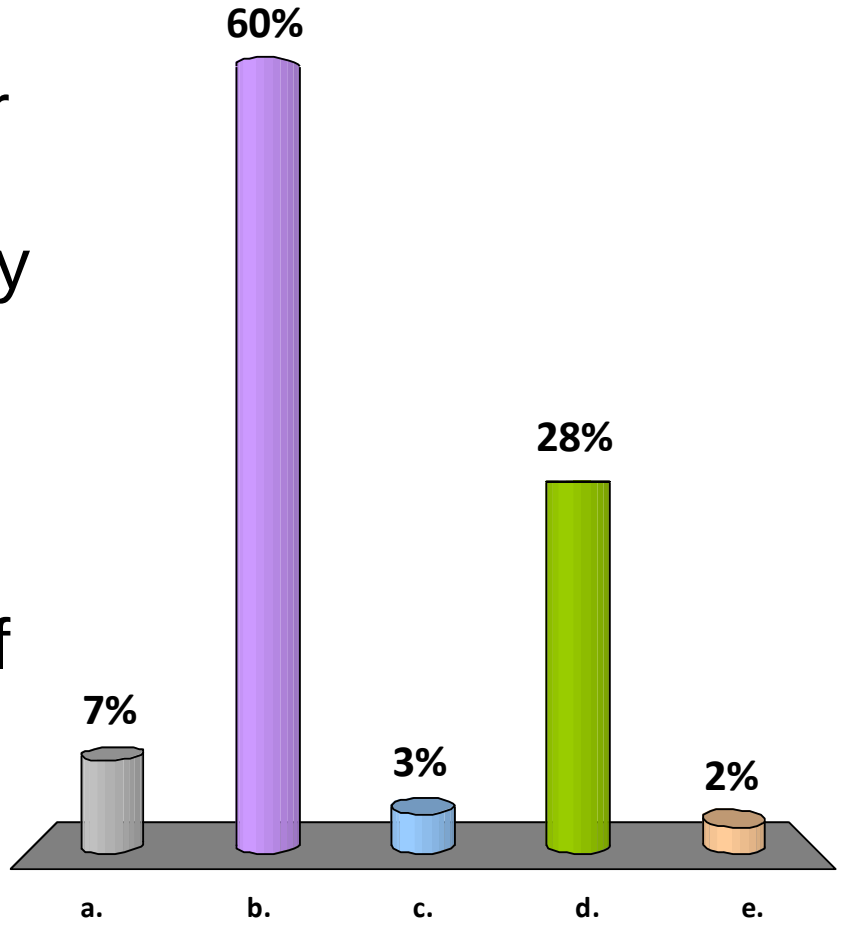




SAMS Question

III-7 Information about the structural detail in an image is:

a. dependent on the Fourier transform.
b. contained in the periphery of *k*-space.
c. blurred by increasing separation of lines in *k*-space.
d. contained in the center of *k*-space.
e. related to gradient duration.



SAMs Q:

III-7 Information about the structural detail in an image is:

a) dependent on the Fourier transform.

b) contained in the periphery of *k*-space.

c) blurred by increasing separation of lines in *k*-space.

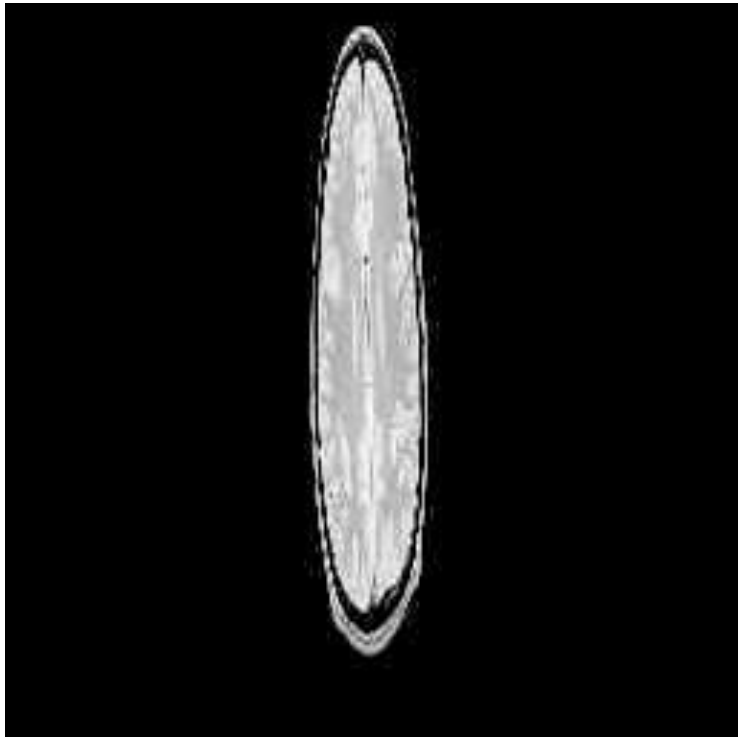
d) contained in the center of *k*-space.

e) related to gradient duration.

Answer: (b).

Ref: "Medical Imaging", A.B. Wolbarst *et al.*, Wiley-Blackwell (2013), p. 337.

Part 4: Spin-Warp; 2D



Spin-Echo Reconstruction (SE vs. FID)
T2 Spin-Relaxation (Static vs. Random Dephasing)
T1-*w*, T2-*w*, and PD-*w* S-E MR Imaging
Spin-Echo / Spin-Warp in 2D (2X2 illustration)

Spin-Echo Reconstruction

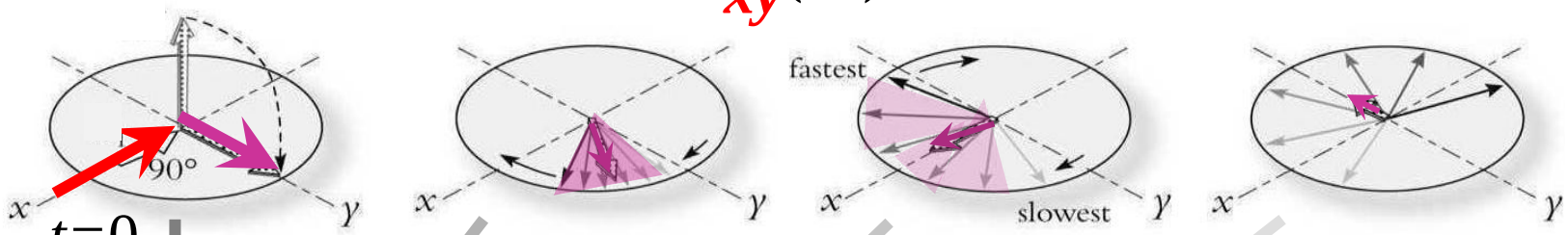
Spin Echo vs. FID

Static Field de-Phasing of FID in x-y Plane of Spins in Each Voxel

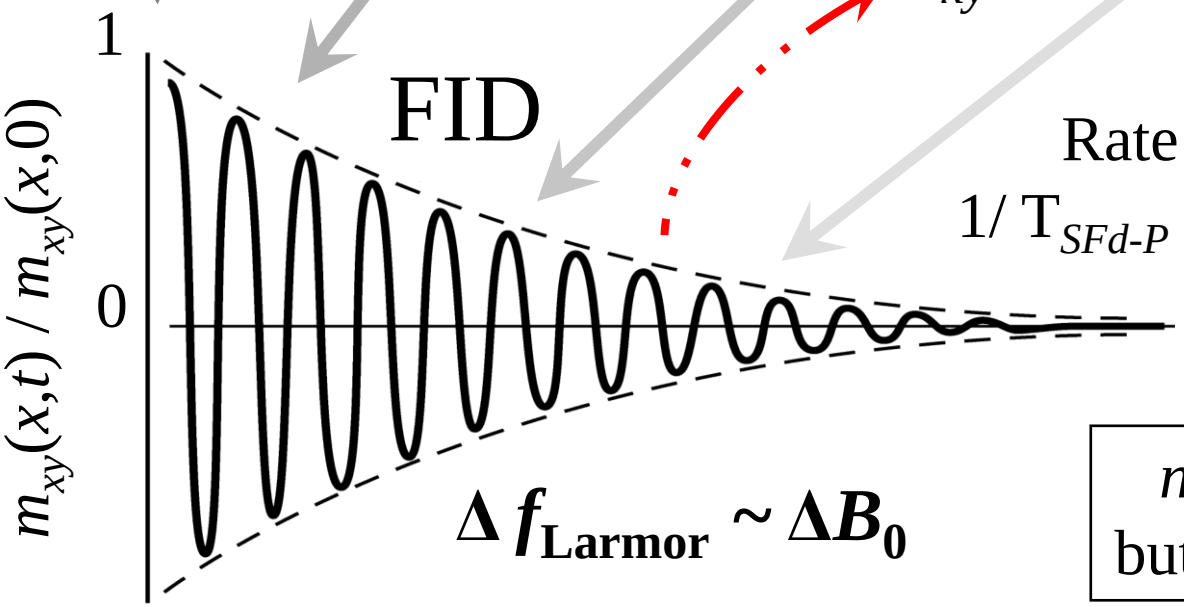
SFd-P from ***static***-field inhomogeneities & susceptibility effects

This de-phasing is ***REVERSABLE!!!!***

$m_{xy}(x,t)$



$m_{xy}(x,t) \sim e^{-t / T_{SFd\text{-}P}}$



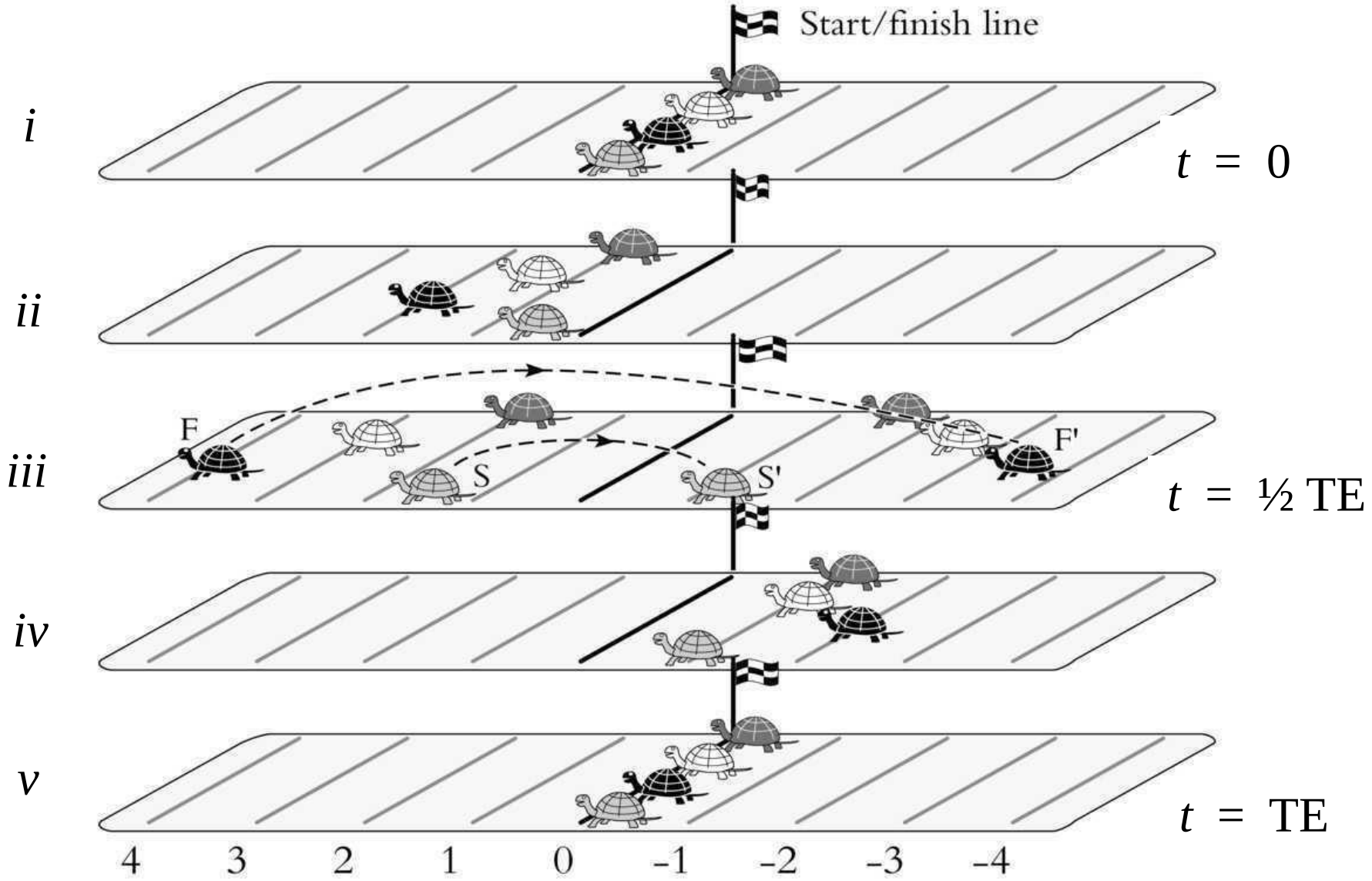
Rate of *SFd-P* decay:

$1/T_{SFd\text{-}P} = \kappa \Delta B_0 + \kappa' \Delta\chi$

t

n.b., $m_{xy}(x,t)$….
but still 1D phantom!

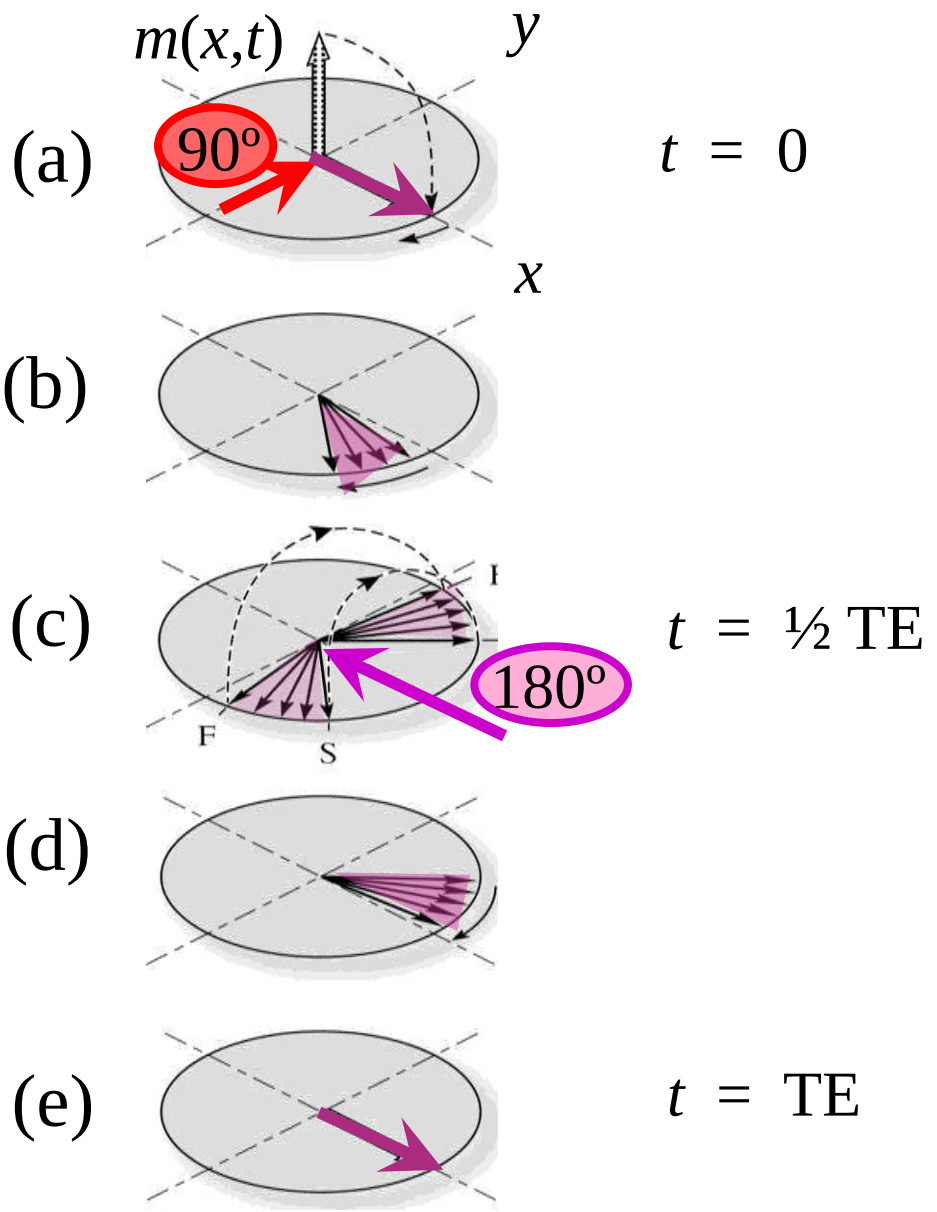
Kentucky Turtle-Teleportation Derby Illustration of Spin-Echo



Spin-Echo Generates an Echo *and* Eliminates Effects of *SFd-P*

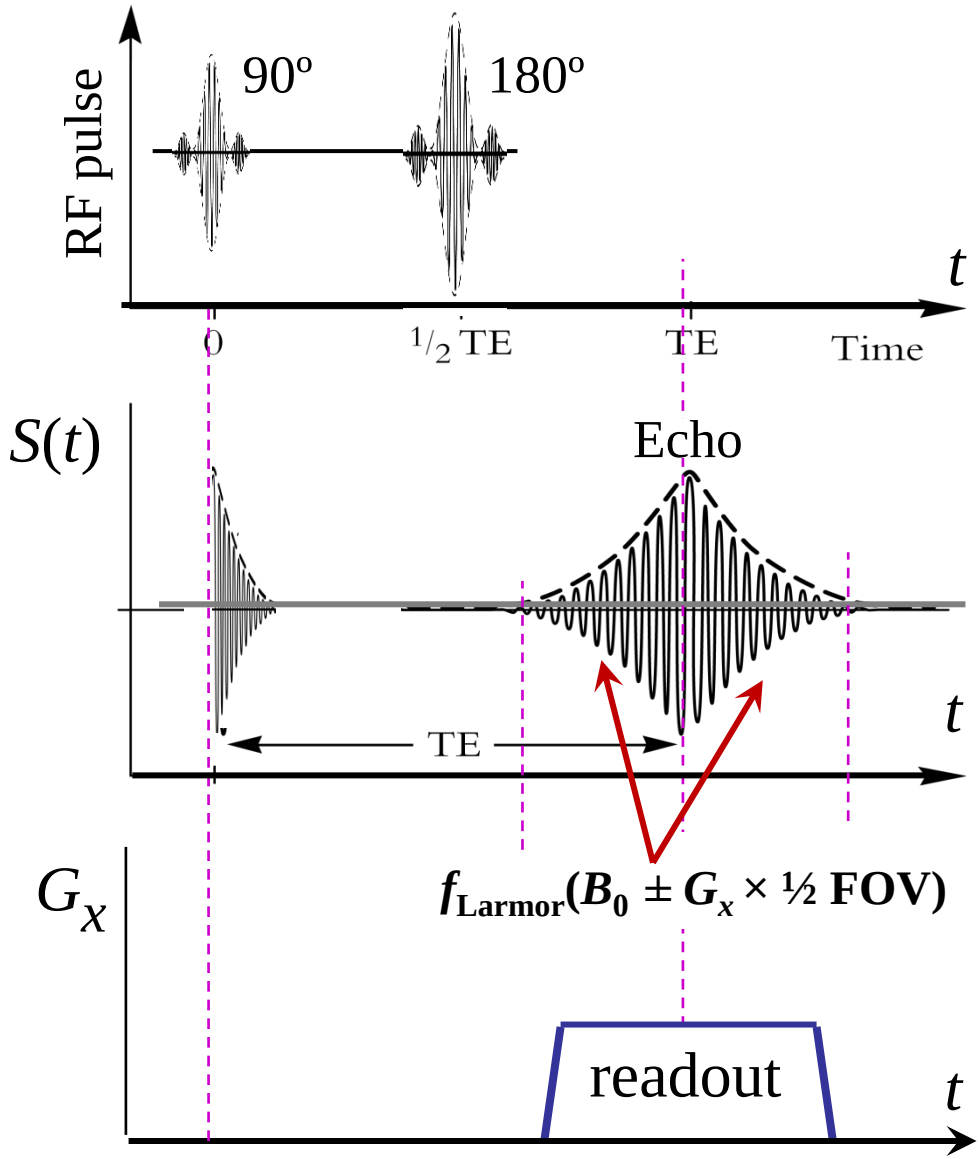
Local magnetic environment for each spin packet is ***static***. So ***this*** de-phasing can be **REVERSED*!!!***

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



S-E Sequence Using 180º RF Pulse and Readout Gradient

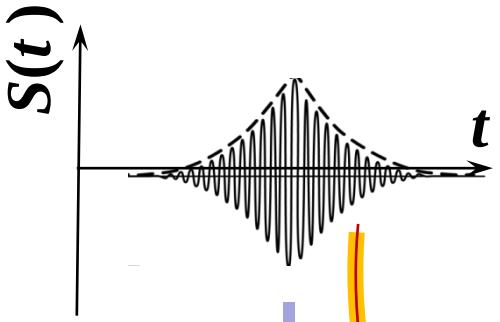
with 256 voxels, e.g., sample/digitize echo signal 256-512 times



k-Space (spatial frequency) Procedure for *SE* in 1D Phantom

$$S(t) = S(k(t)) \sim \int m(x)\, e^{-2\pi i\, k_x(t)\, x}\, dx$$

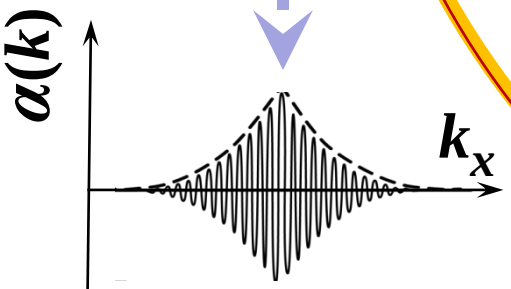
Sampled **SE** signal



$S(t)$ made up of contributions from $m(x)$ for all x;
$\boldsymbol{m}(x)$ can be extracted from $S(t)$ with $k_x \rightarrow x$ $\mathcal{F}$:

$$k_x = (\gamma/2\pi)\, G_x\, t$$

$$m(\boldsymbol{x}) = \int S(t)\, e^{-2\pi i\, \boldsymbol{k}(t)\cdot\boldsymbol{x}}\, d^3k$$



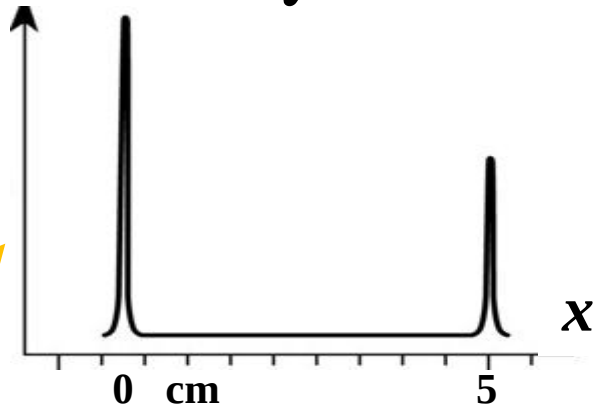
k-space representation

$\alpha(k)$: Amplitude and phase of spatial frequency (k)

$\mathcal{F}^{-1}$

$k_x \rightarrow x$

$$R(x) = \int \alpha(k)\, e^{+ikx}\, dk$$



Real-space representation

During Readout, k_x Increases Linearly with t

$$S(t) = S(k(t)) \sim \int m(x)\, e^{-2\pi i\, k_x(t)\, x}\, dx \ .$$

$k_x(t)$ for all voxels increases linearly with t while the echo signal is being received and read. With signal sampled at 512 adjacent instances spaced Δt apart, t represents the exact sampling time: the n^{th} sample is taken a time $t_n = n \times \Delta t$ after G_x was turned on. Different values of k correspond to different sampling times. In particular, for t_n,

$$k_n = [G_x \gamma / 2\pi]\, t_n$$

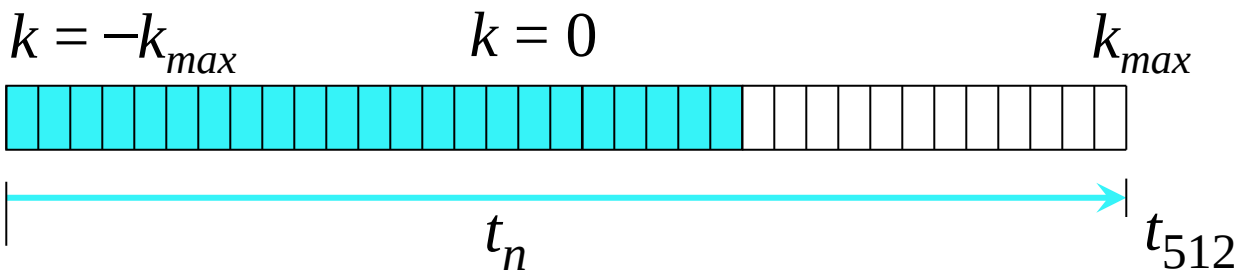
Larger *k*-values correspond to **greater spatial frequencies**!

During Readout, k_x Increases Linearly with t

Signal is sampled sequentially at 256-512 times spaced Δt apart; t_n *i*s the exact sampling time 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 (and larger k-values) correspond to greater spatial frequencies.

S-E reads out from $k_n = -k_{max}$ to $k = 0$ to $k = k_{max}$



T2 Spin-Relaxation

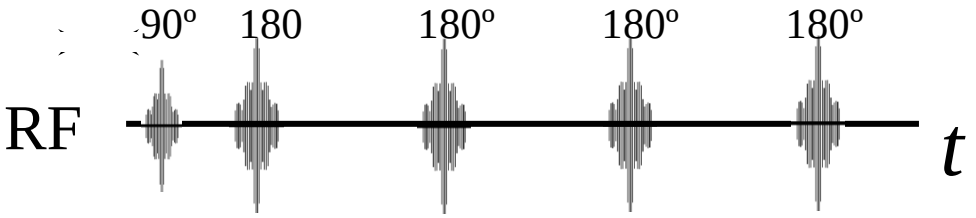
T2 Relaxation refers to the rate at which the transverse magnetization decays (disappears).

T2 relaxation results from T1-Events and
Non-Static, Random, Non-Reversible
Proton-Proton Dipole Interactions
Both Contribute to the decay rate (1/T2)

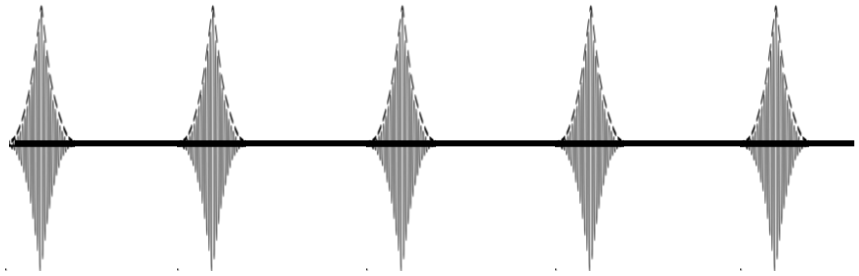
(Specifically, any process that either reduces the number of transverse spins or their relative phase relationship will add to T2 relaxation.)

Thus, the Spin-Echo Signal Intensity $S(t)$ ***Does*** Decay, and decays much Faster than 1/T1!

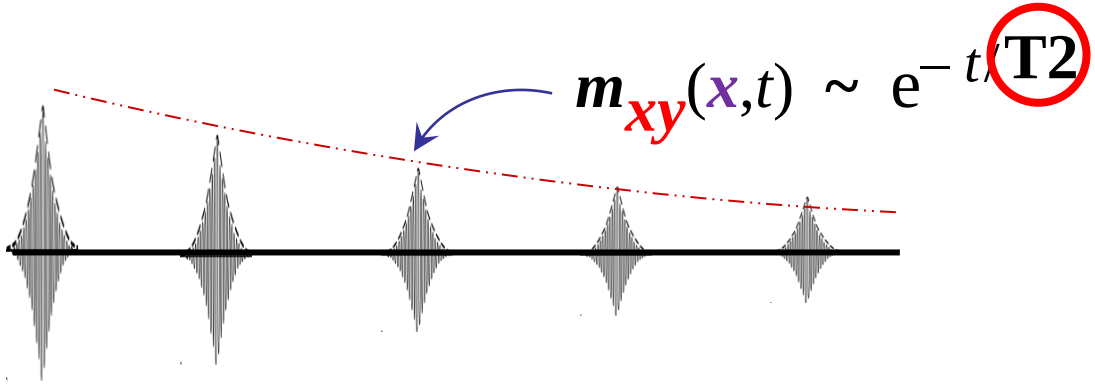
Spin-Echo Spin De-Phasing Is Caused by T1 Events *as well as* By ***Random*****,** Proton-Proton Dipole Interactions



Spin Multi-Echo:
we ***might*** *expect* $S(t)$

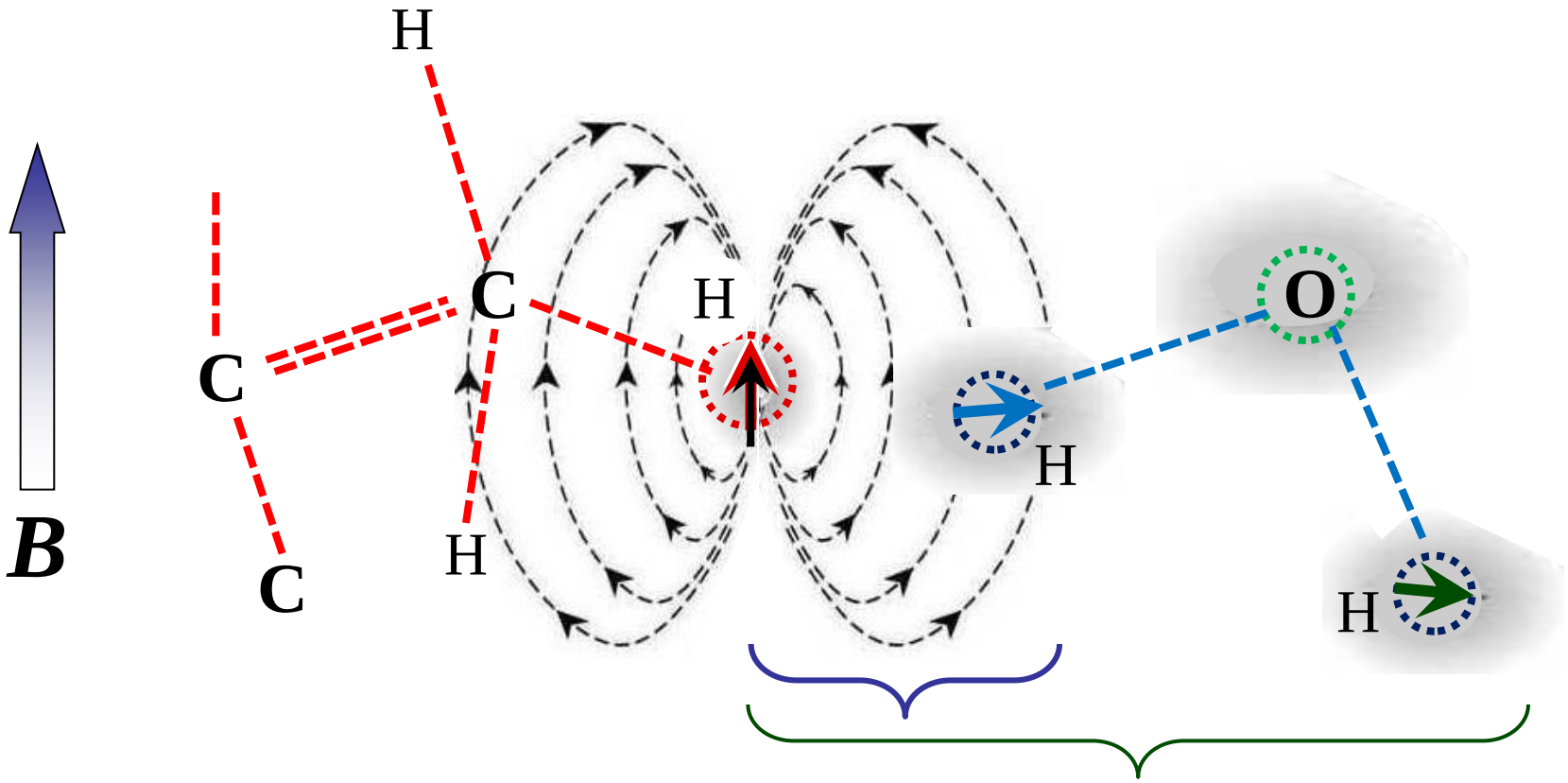


However, Harsh Reality



Spin-Spin (Secular) De-Phasing in Bound Water

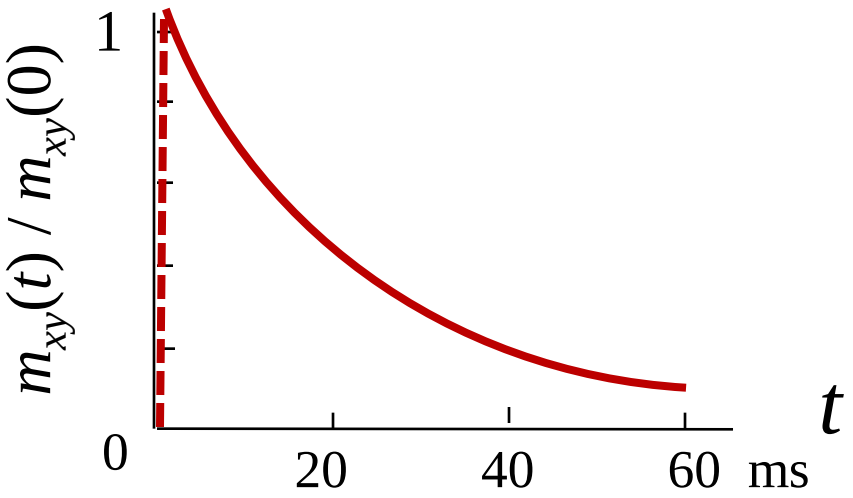
with its protons precessing in the *x-y* plane



Quasi-static *spin-spin interactions*

does ***not*** involve exchange of energy like T1 relaxation!

Exponential T2-Caused De-Phasing of $\boldsymbol{m}_{xy}(t)$ in x-y Plane



rate

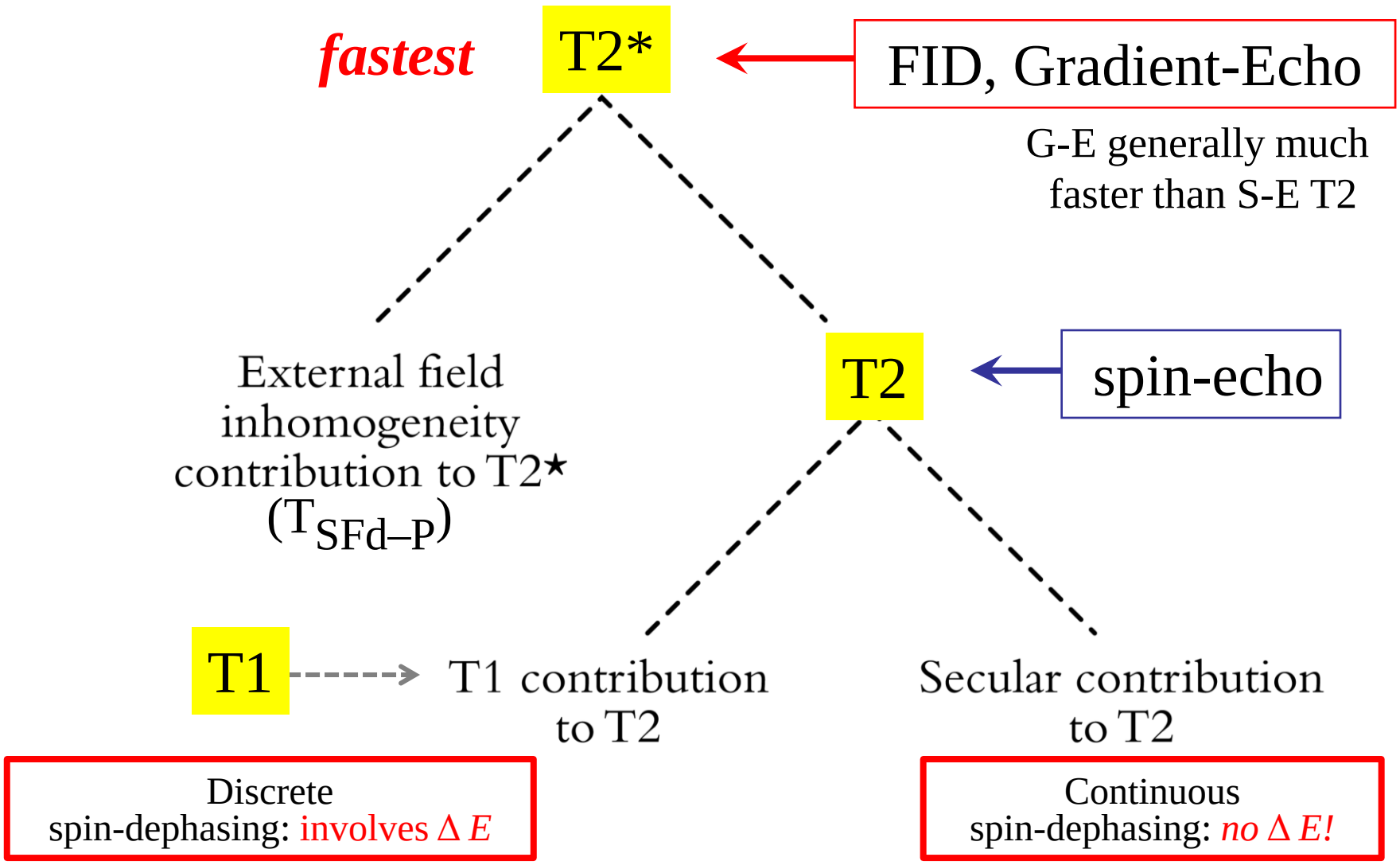
$$d\boldsymbol{m}_{xy}(t)/dt = -(\mathbf{1/T2})\,\boldsymbol{m}_{xy}(t)$$

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

$$d\boldsymbol{m}(x,t)/dt = \gamma\,\boldsymbol{m}(x,t)\times\boldsymbol{B}_{\boldsymbol{z}}(x) - \frac{[m(x,t)-m_{\mathbf{0}}(x)]\hat{\boldsymbol{z}}}{T1} - \frac{[m_x\,\hat{\boldsymbol{x}} + m_y\,\hat{\boldsymbol{y}}]}{T2}$$

classical Bloch Equation

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



T2 relaxation refers to:

7%

a. The rate at which longitudinal magnetization recovers.

11%

b. The rate at which longitudinal magnetization disappears.

7%

c. The rate at which transverse magnetization recovers.

72%

d. The rate at which transverse magnetization disappears.

3%

e. The rate at which tissue is magnetized.

T2 relaxation refers to:

a) the rate at which longitudinal magnetization recovers.
b) the rate at which longitudinal magnetization disappears.
c) the rate at which transverse magnetization recovers.
d) the rate at which transverse magnetization disappears.
e) the rate at which tissue is magnetized.

Answer: (d).

Ref: "Medical Imaging", A.B. Wolbarst *et al.*, Wiley-Blackwell (2013), p. 364.

Tissue Contrast Weighting (*w*) in S-E MRI
T1-*w*, T2-*w*, and PD-*w*

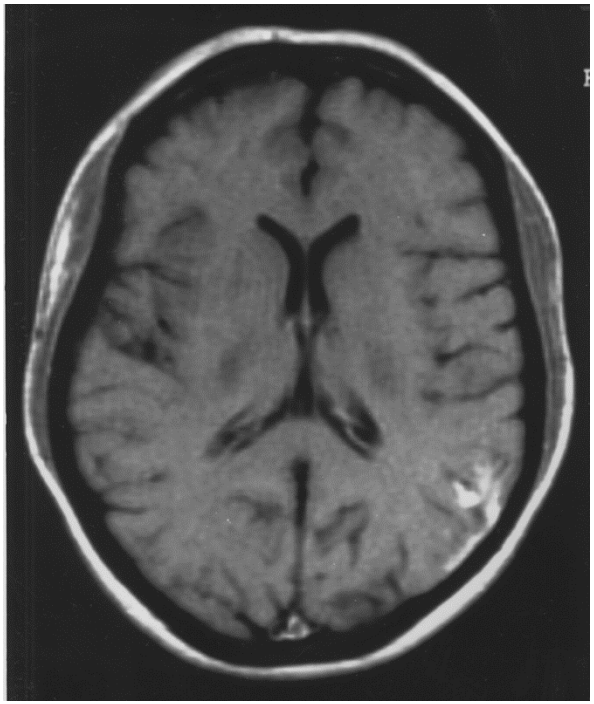
Typical T1 and T2 Relaxation Times

Relaxation Rates: 1/T2 ~ 10 × (1/T1)

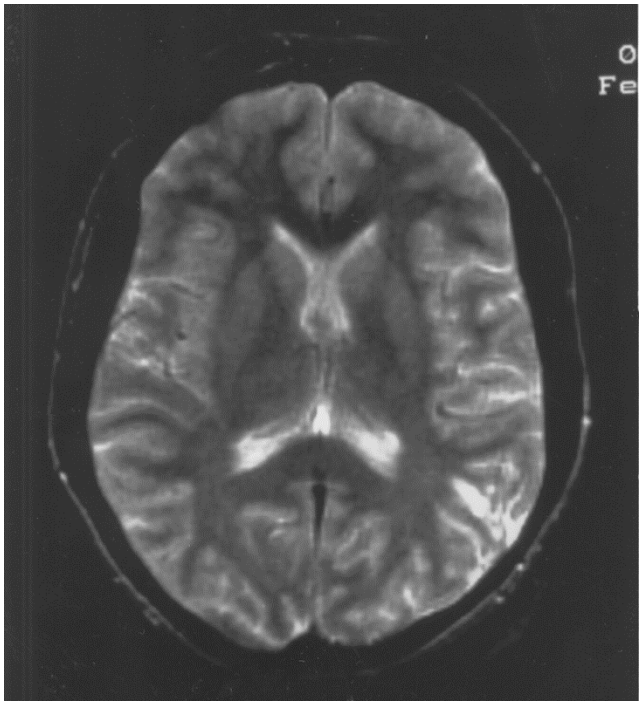
Tissue	PD p^+/mm^3, rel.	T1, *1 T* (ms)	T1, *1.5 T* (ms)	T1, *3 T* (ms)	T2 (ms)
pure H_20	1	4000		4000	4000
brain					
CSF	0.95	2500	2500	2500	200
white matter	0.6	700	800	850	90
gray matter	0.7	800	900	1300	100
edema			1100		110
glioma		930	1000		110
liver			500		40
hepatoma			1100		85
muscle	0.9	700	900	1800	45
adipose	0.95	240	260		60

Three Different Forms of MRI Contrast

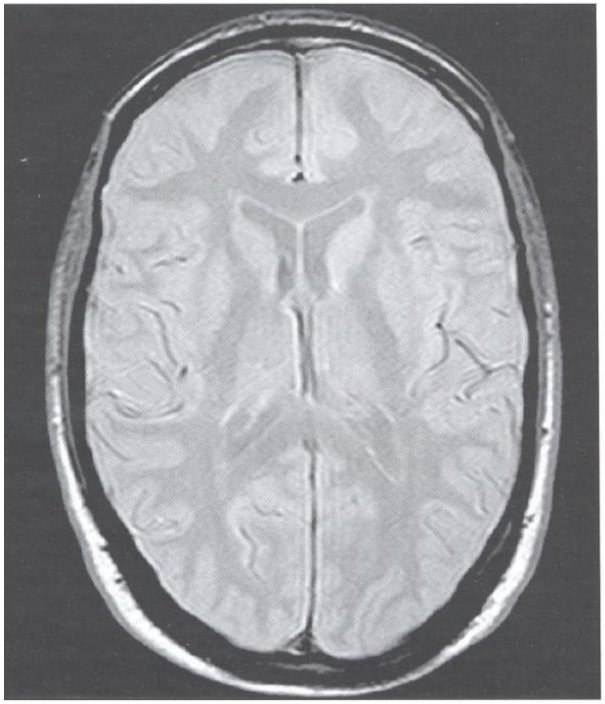
created by, and reflecting, three quite different physical properties



T1

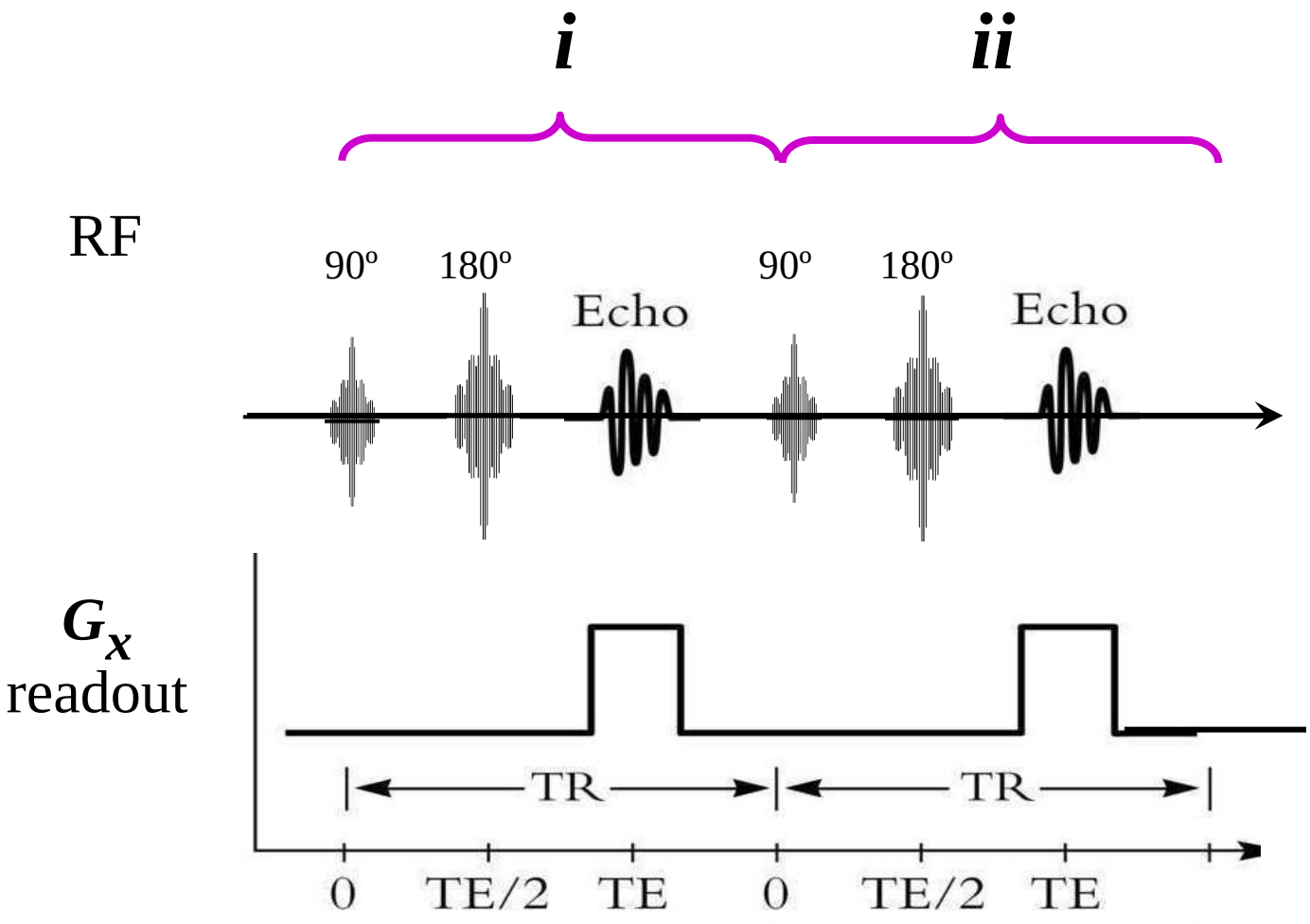


T2



PD

Multiple Spin-Echo Pulse Sequence



MRI Signal Strength at t = TE Depends on….

Inherent Parameters	Operator Settings
T1	TR
T2	TE
PD	

time since previous S-E sequence

$$S(t = \mathrm{TE}) \sim \mathrm{PD}\,(1 - e^{-\mathrm{TR}/\mathrm{T1}})\, e^{-\mathrm{TE}/\mathrm{T2}}$$

proton density

prior regrowth along z-axis

current dephasing in x-y plane

bright regions:

m_{xy} large at t = TE

In a spin-echo pulse sequence, a short TE will:

13%	a.	**Minimize T1 image contrast**
83%	b.	**Eliminate T2 effects**
1%	c.	**Result in a lower SNR**
2%	d.	**Increase the effects of static-field de-phasing**
2%	e.	**Enhance tissue susceptibility differences**

In a spin-echo pulse sequence, a short TE will:

(a) Minimize T1 image contrast
(b) Eliminate T2 effects
(c) Result in a lower SNR
(d) Increase the effects of static-field dephasing
(e) Enhance tissue susceptibility differences

Answer: (b).

Ref: "Medical Imaging", A.B. Wolbarst *et al.*, Wiley-Blackwell (2013), p. 368.

T1-*weighted* – Short TE to Eliminate T2 Contribution

TR ~ $T1_{av}$ to maximize contrast

Short TE to minimize T2 impact

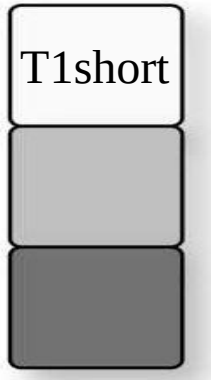
$$S(t \sim TE) \sim PD\,(1 - e^{-TR/T1})\,e^{-TE/T2}$$

proton density

prior regrowth along *z*-axis

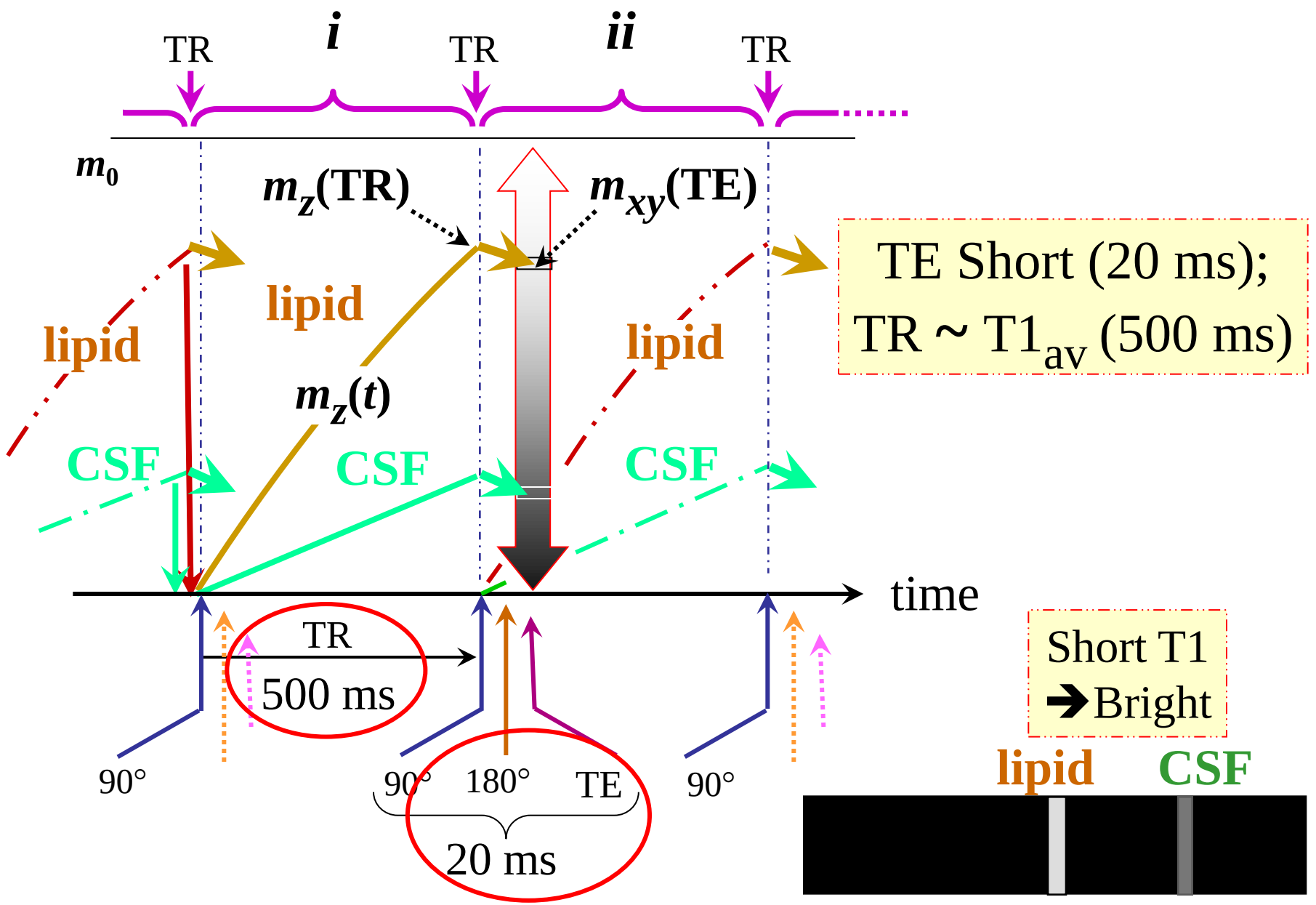
current de-phasing in *x*-*y* plane

T1-*w*



Short-T1 tissues bright
(sprightly brightly)

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



T2-*w* – Long TR to Eliminate T1 Contribution

For T2-*w* imaging

Long TR: eliminate T1 impact

TE at mid-T2: maximize contrast

$$S(t \sim \mathrm{TE}) \sim \mathrm{PD}\,(1 - e^{-\mathrm{TR}/\mathrm{T1}})\, e^{-\mathrm{TE}/\mathrm{T2}}$$

proton density

prior regrowth along *z*-axis

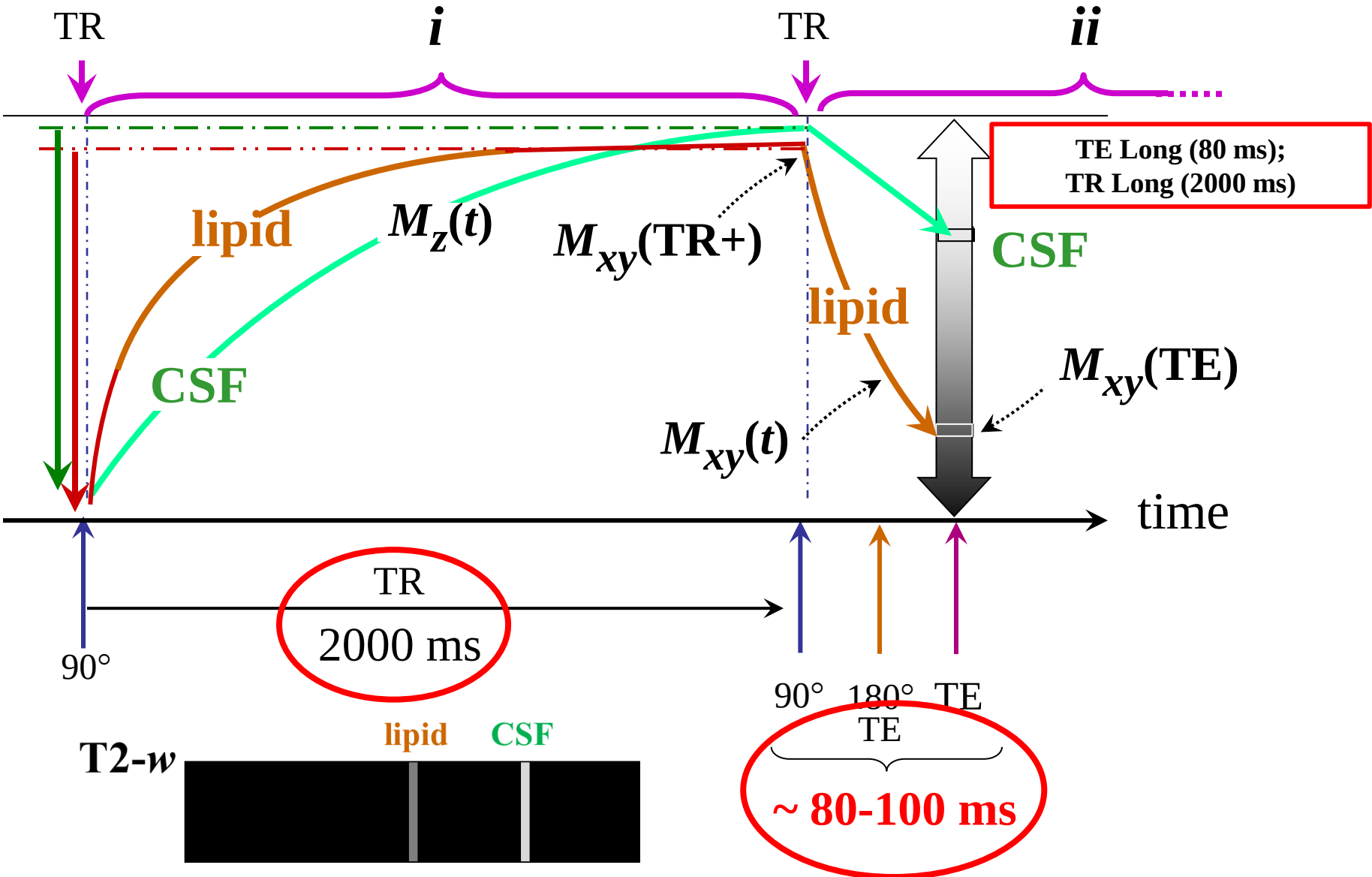
current de-phasing in *x*-*y* plane

T2-*w*

Long-T2 tissues bright

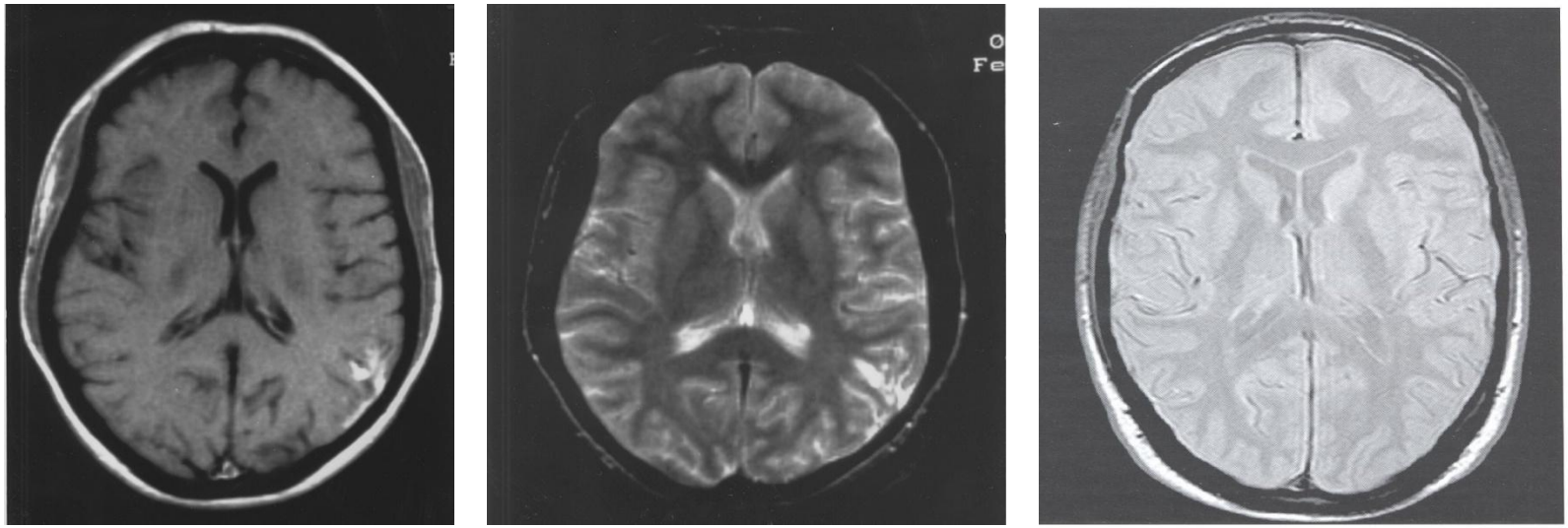
T2-*w* – Long TR to Eliminate T1 Contribution

TE ~ mid-T2 to maximize T2 contrast



PD-, T1-, & T2-Weighted Spin-Echo Images (1.5T)

T1-*w* T2-*w* PD-*w*



TR (ms)	mid- (~$T1_{av}$) 300 – 700	long 1,500 – 3,500	long 1,500 – 3,500
TE (ms)	short 0 – 25	mid- (~$T2_{av}$) 60 – 150	short 10 – 25
Bright	short T1	long T2	high PD
SNR	good	lower	best

Which of the following would appear bright on a T2-weighted image of the brain?

87%	a. CSF
7%	b. Fat
0%	c. Bone
6%	d. White Matter
0%	e. Air

Which of the following would appear bright on a T2-weighted image of the brain?

(a) CSF
(b) Fat
(c) Bone
(d) White matter
(e) Air

Answer: (a) cerebral spinal fluid.

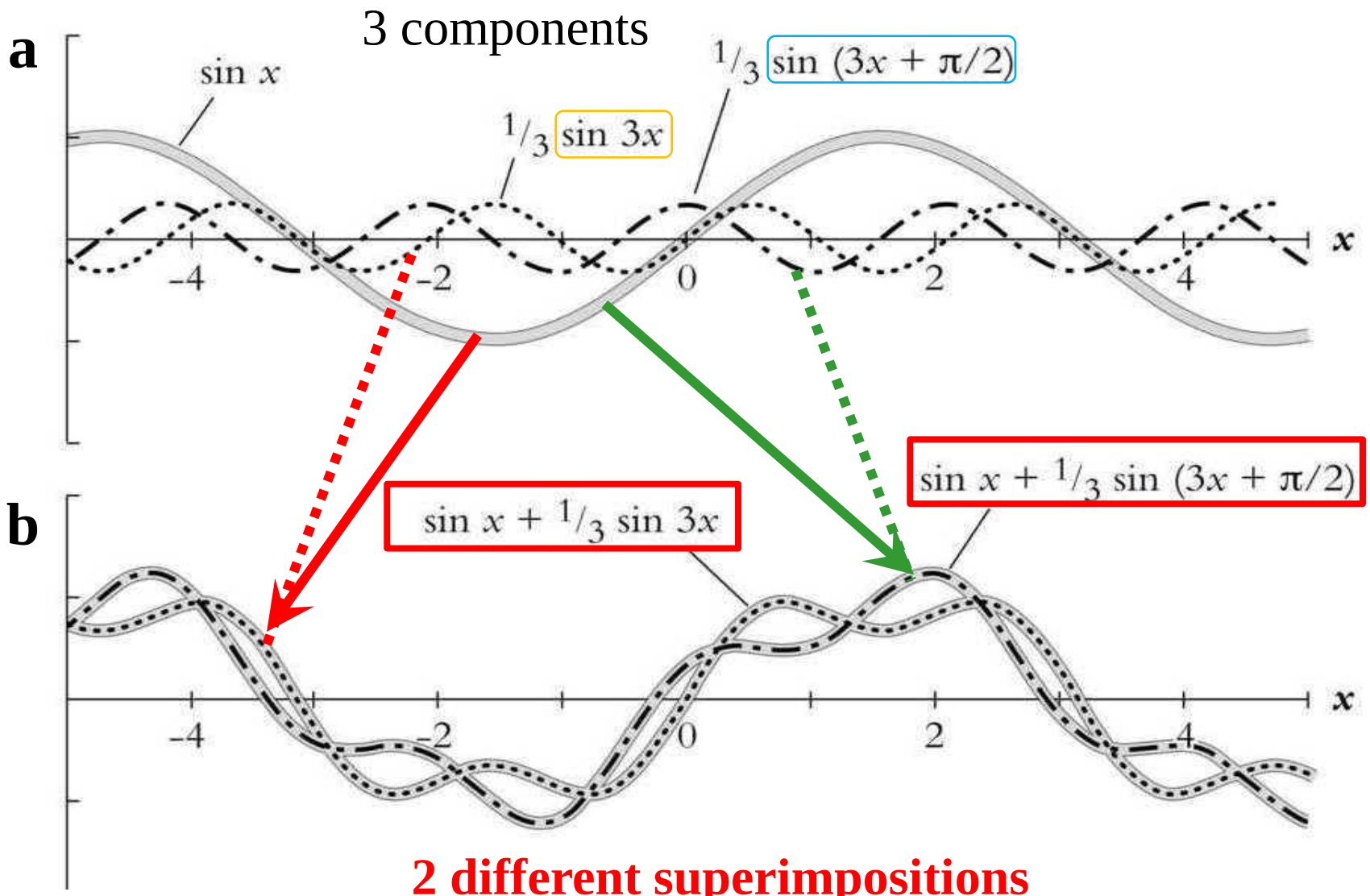
Ref: "Medical Imaging", A.B. Wolbarst *et al.*, Wiley-Blackwell (2013), p. 368.

2D Images using Spin-Echo/Spin-Warp

One of the earliest 2D imaging methods was Sensitive Point Reconstruction.

This method used oscillating gradient fields to produce a net field stable in only one voxel at a time. That “sensitive point” was then scanned in space.

In 2-D MRI, phase is as important as frequency and is used to add a second spatial dimension.

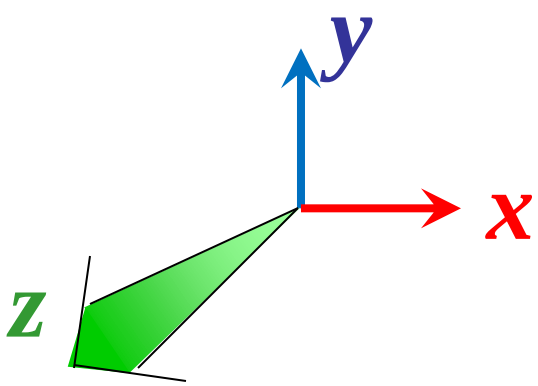


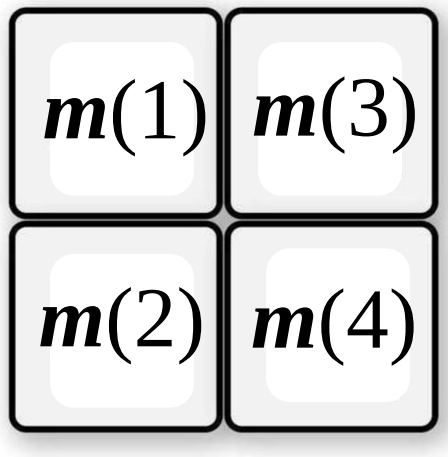
2 different superimpositions

2D Spin-Warp 2X2 Matrix Illustration

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

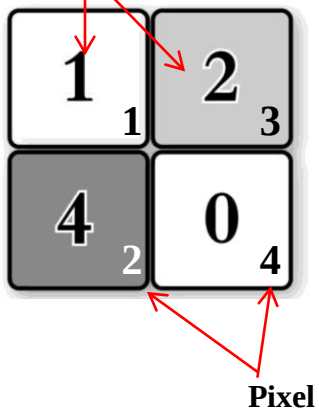
after 90º pulse drives $M(t)$ into x-y plane…





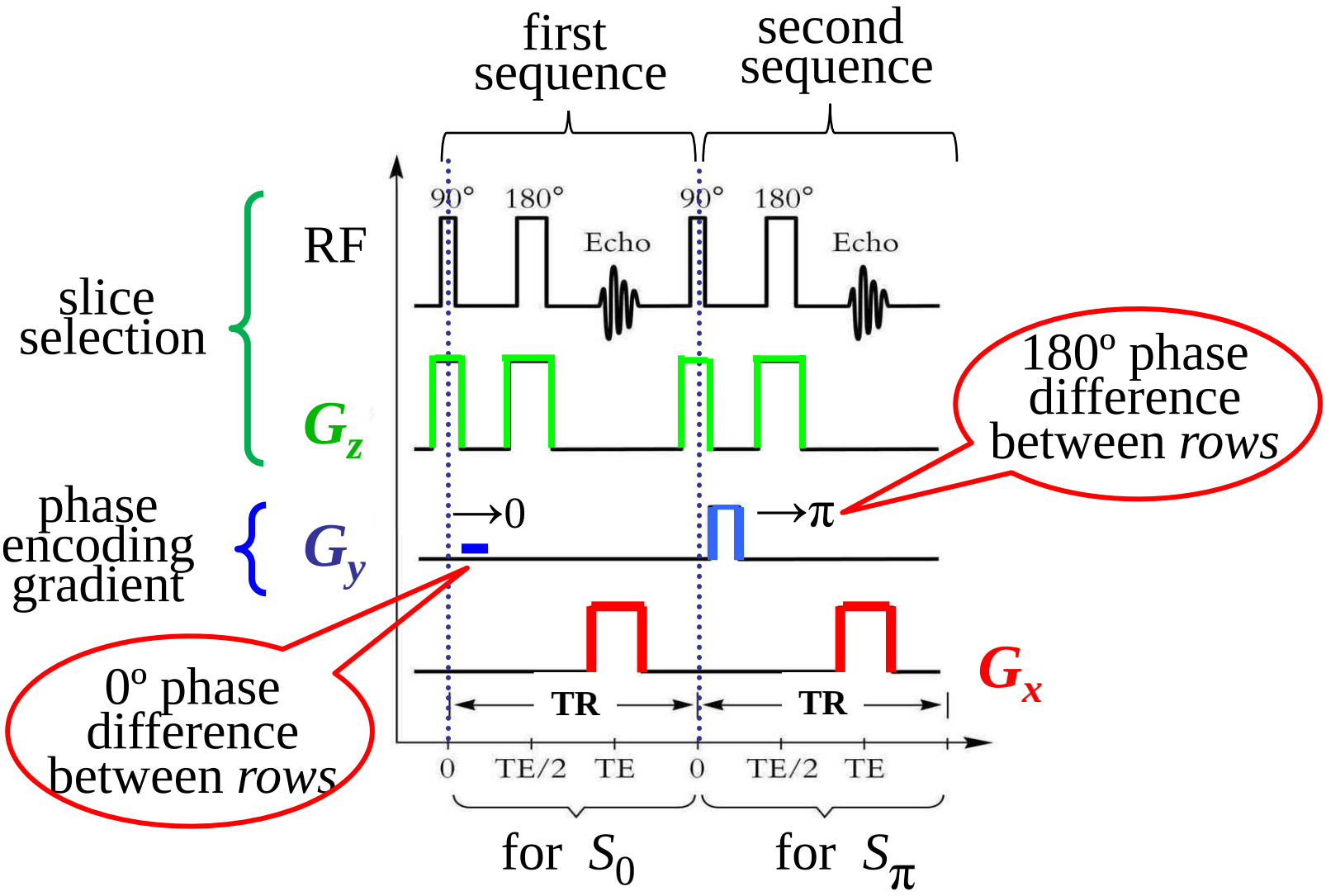
viewed from feet to head

PD Map



Spin-Echo, Spin-Warp Sequence for 2×2 Matrix

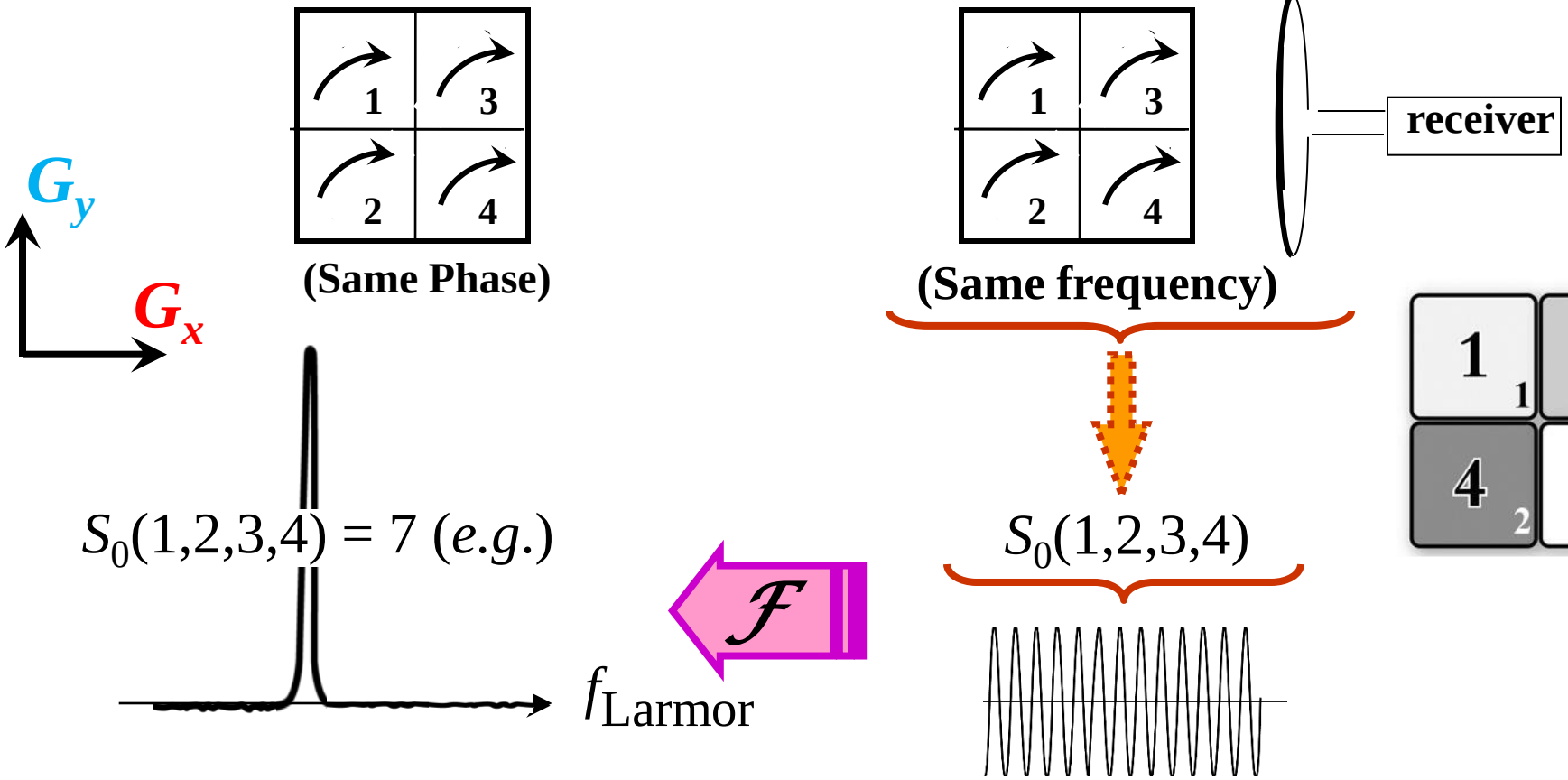
involves 2 spin-echo pulse sequences



e.g., **No** *x*- or *y*-Gradients: $G_y = G_x = 0$

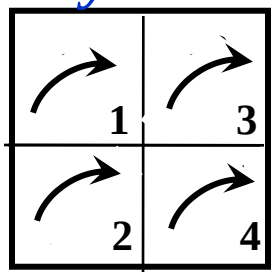
After (G_y=0) phase encoding

during (G_x= 0) readout

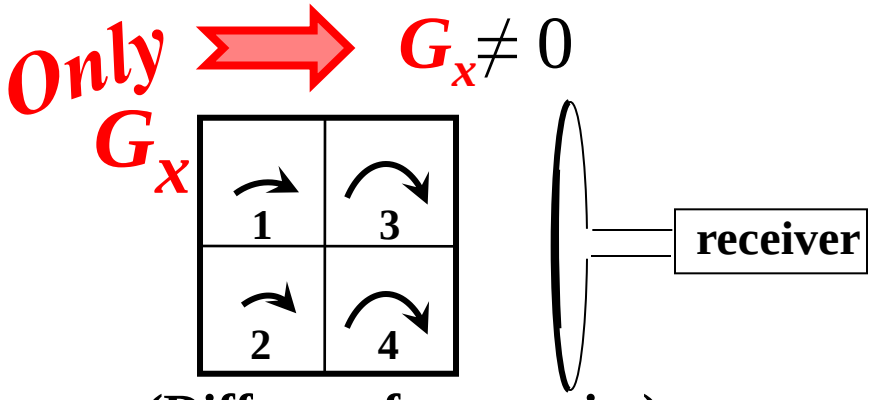


$$\mathcal{F}[S_0(1,2,3,4)] \rightarrow 7 = m(1) + m(2) + m(3) + m(4)$$

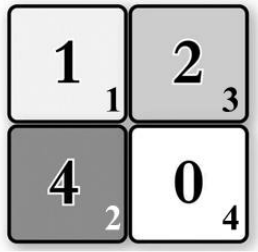
1st S-E Sequence: *x*-Gradient Only, On During Readout

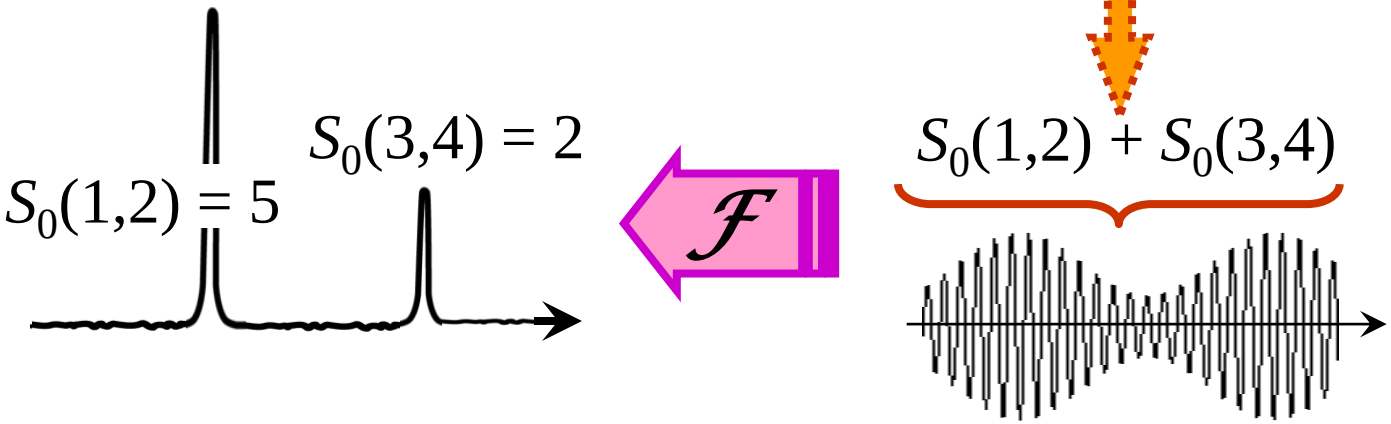


(Same phase)



(Different frequencies)

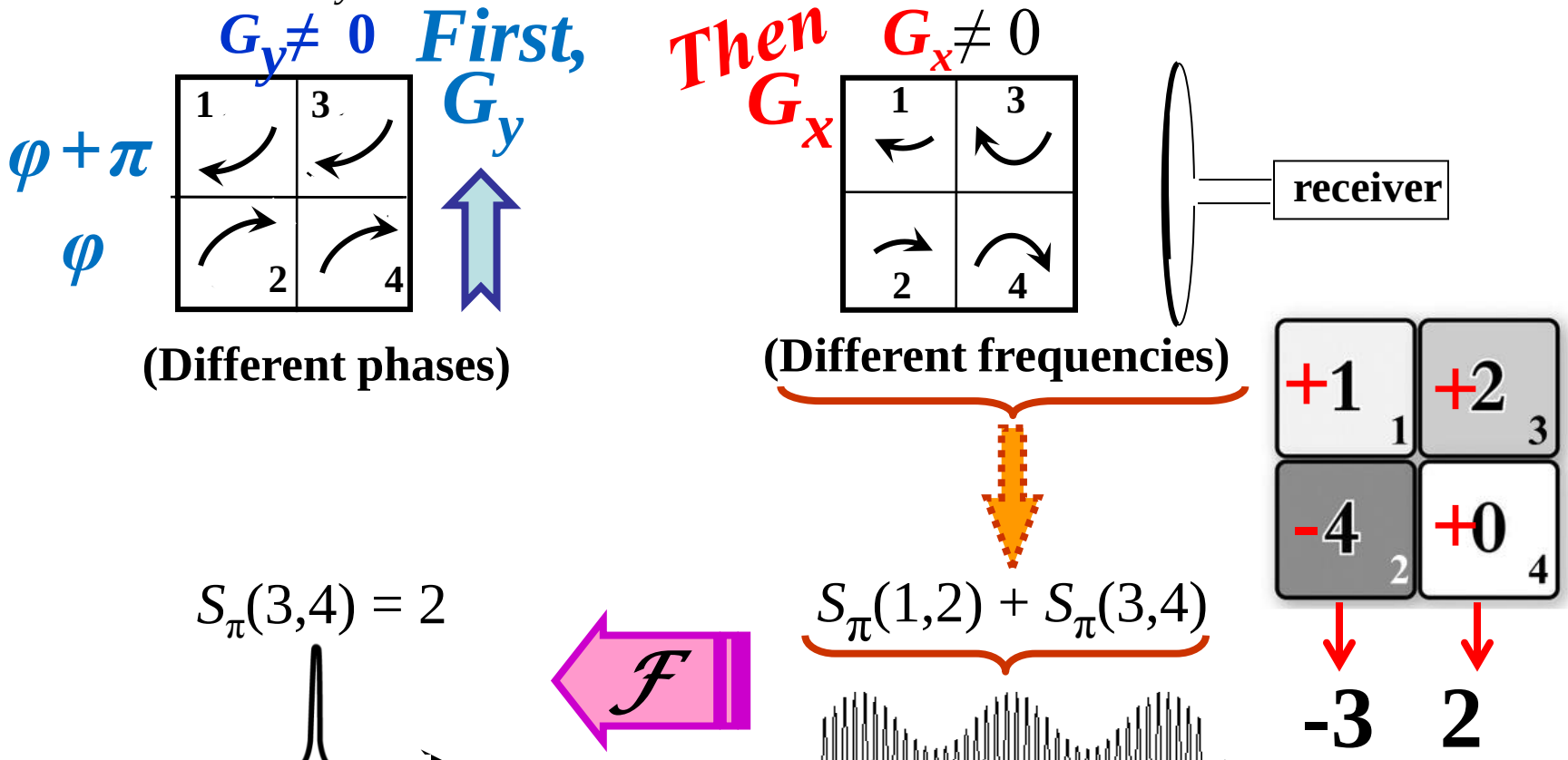




$$\mathcal{F}[S_0(1,2) + S_0(3,4)] \rightleftarrows \begin{matrix} S_0(1,2) = 5 = m(1) + m(2) \\ S_0(3,4) = 2 = m(3) + m(4) \end{matrix}$$

2nd S-E Sequence: *y*-Gradient, then *x*-Gradient

$G_y \rightarrow 180^\circ\,\Delta$. G_x On During Readout



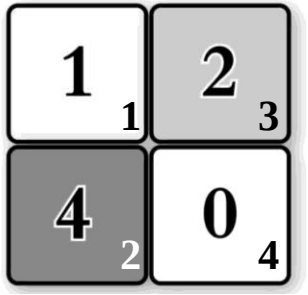
4 Equations in 4 Unknowns

in phase: add

$$S_0(1,2) + S_0(3,4) \xrightarrow{\mathcal{F}} \begin{cases} S_0(1,2) = 5 = m(1) + m(2) \\ S_0(3,4) = 2 = m(3) + m(4) \end{cases}$$

180° out of phase: subtract

$$S_\pi(1,2) + S_\pi(3,4) \xrightarrow{\mathcal{F}} \begin{cases} S_\pi(1,2) = -3 = m(1) - m(2) \\ S_\pi(3,4) = 2 = m(3) - m(4) \end{cases}$$



Solution:

$m(1) = 1$

$m(2) = 4$

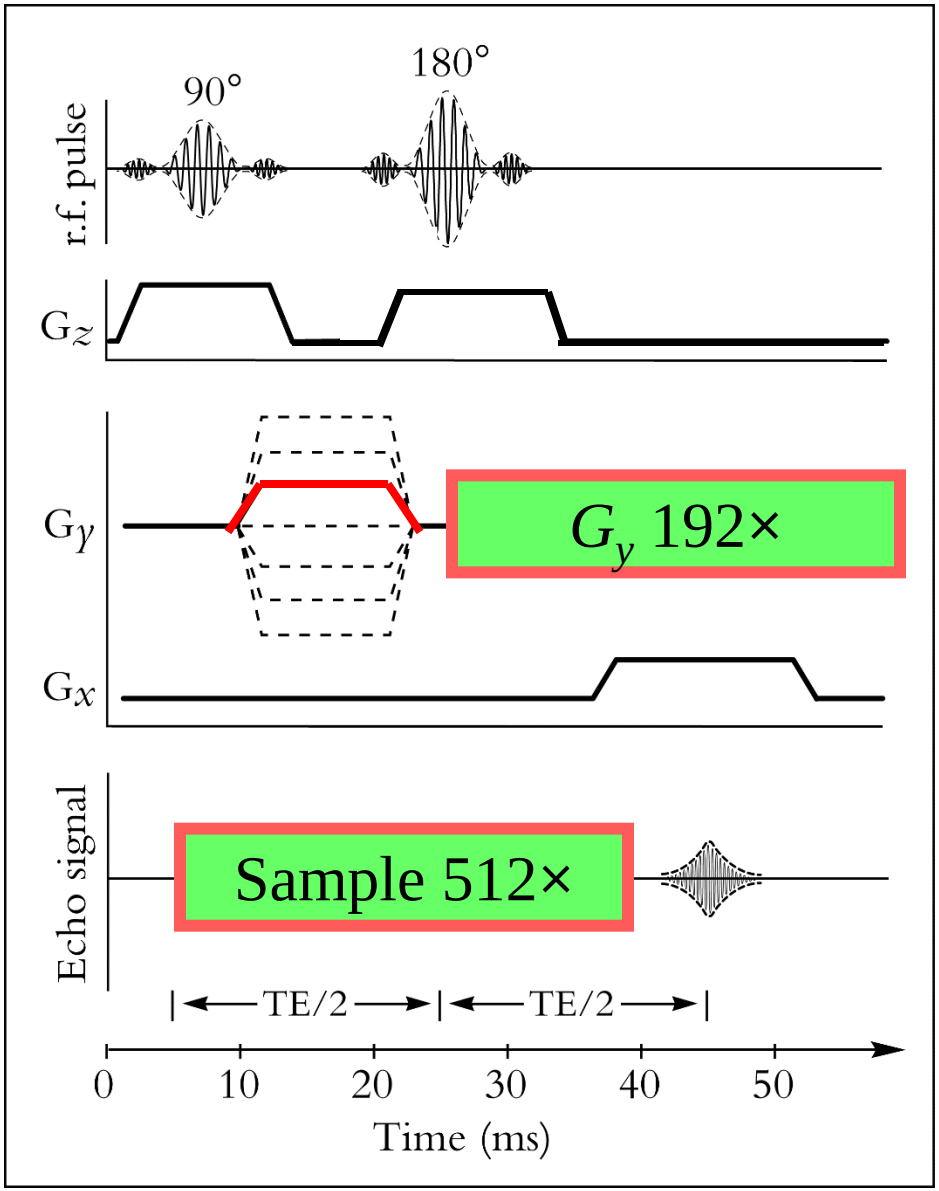
$m(3) = 2$

$m(4) = 0$

check:

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

Spin-Echo, Spin-Warp…. (256×192 matrix)



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

Free precession with G_y on phase-encodes y-position

$G_y(n) = n \times G_y(1)$, $n = 1, 2, \ldots, 192$

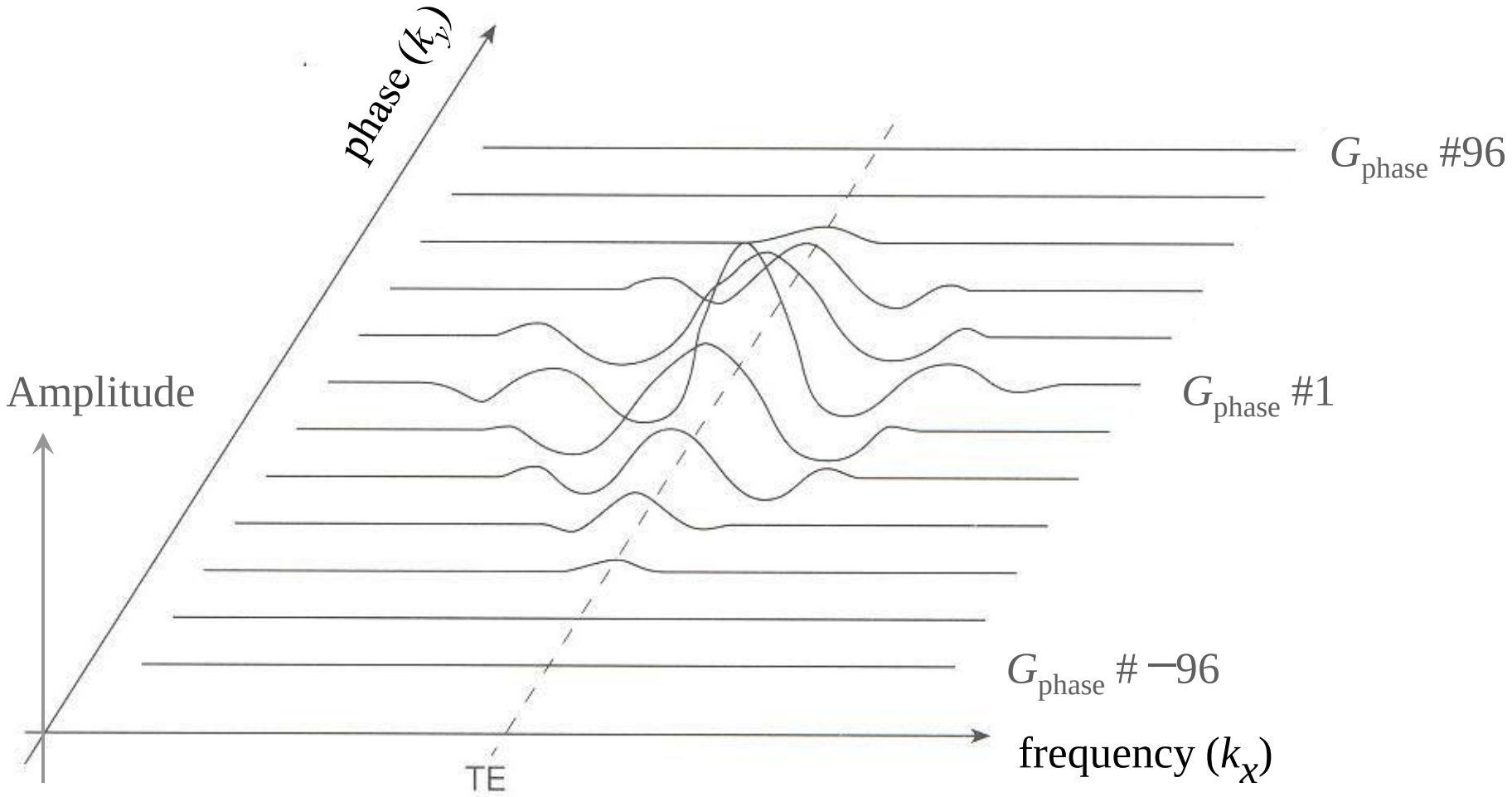
Reading NMR signal with G_x on frequency-encodes x-position

Repeat, but with $G_y(n+1)$

$n = 1, 2, \ldots, 192$

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

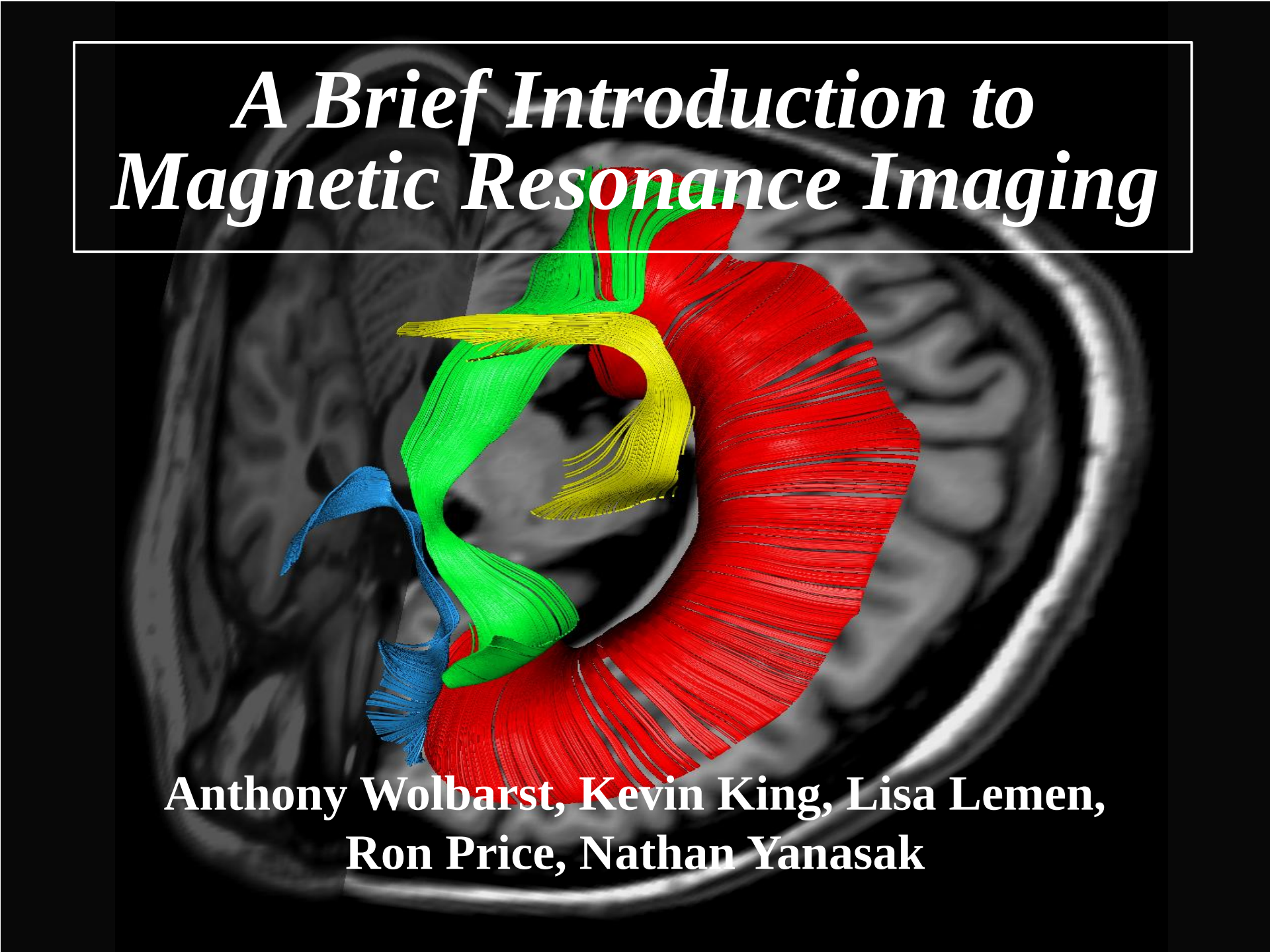
each line in *k*-space from data obtained during one M_{xy}(TE) readout;
different lines for different phase-encode gradient strength



Multiple Important MR 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

A Brief Introduction to Magnetic Resonance Imaging



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