The Principles of Quantitative MRI

Geoffrey D. Clarke
Dept. of Radiology

University of Texas Health Science Center at San Antonio

Overview

- Excitation & Signal Collection Process
- Gradients for Spatial Localization
- Measuring Tissue Volumes
- Measuring NMR Properties of Tissues
- Tissue Physiology Measurements
- ~~Tissue Biochemistry Measurements~~

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RF Nonuniformities

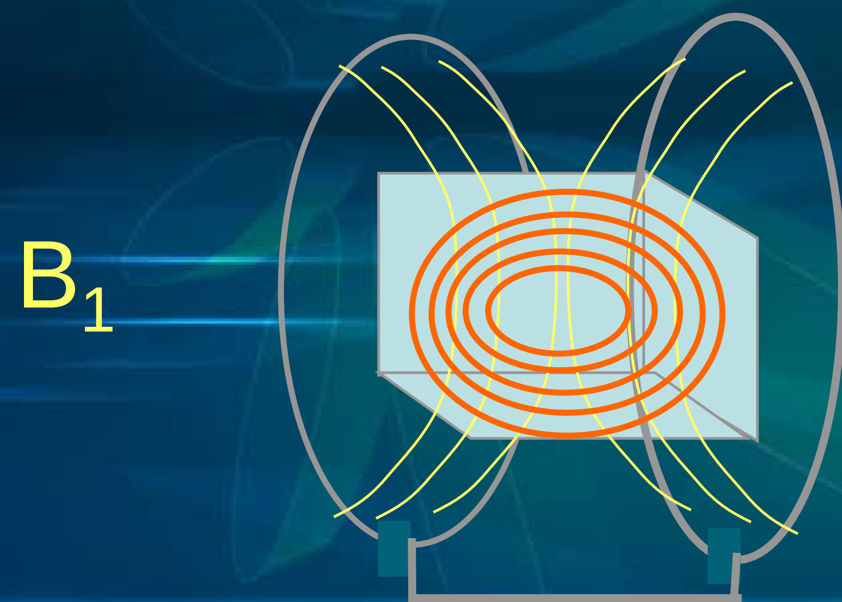
- **Radio Frequency Field Nonuniformities are the <u>Single Biggest</u> Cause of Errors in qMRI**
- **RF Nonuniformities Increase as the B_o-Field Increases**
- **Dielectric Resonance Effects Become Pronounced at High B_o**

B1 TX Field

- Directly related to current in TX coil
  - Depends on Q of coil & coil loading
- Depends on TX Coil Geometry
- TX power auto-adjusted (pre-scan)
  - Values should be know to 1%
  - 1% = 0.086 dB
- TX nonlinearities
- RF pulse droop

NMR Signal

$$\delta v(t) = \omega_o B_{1x,y} M_{x,y} \delta V_s \cos(\omega_o t)$$



RF Pulse Bandwidth (BW)

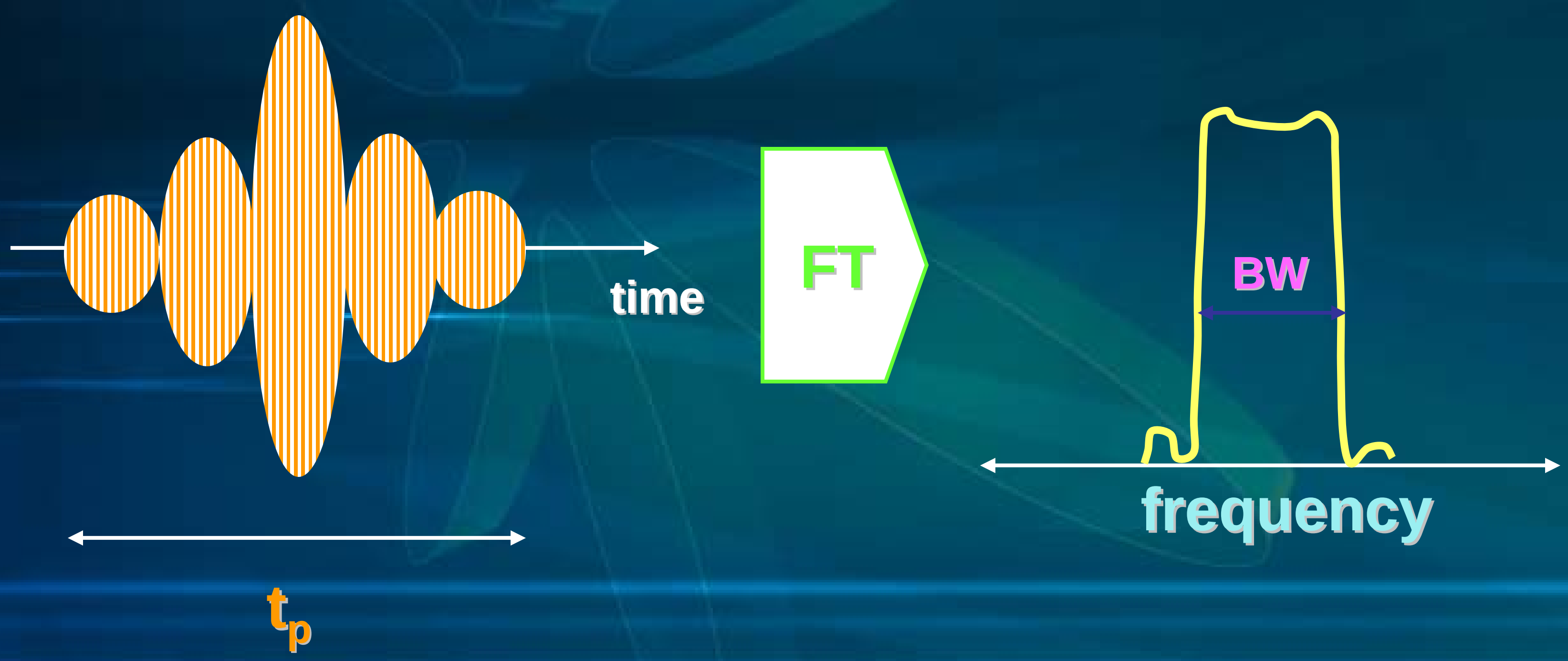
BW is inversely proportional to RF pulse duration:

Slice select gradient is scaled based on BW & G_{ss}:

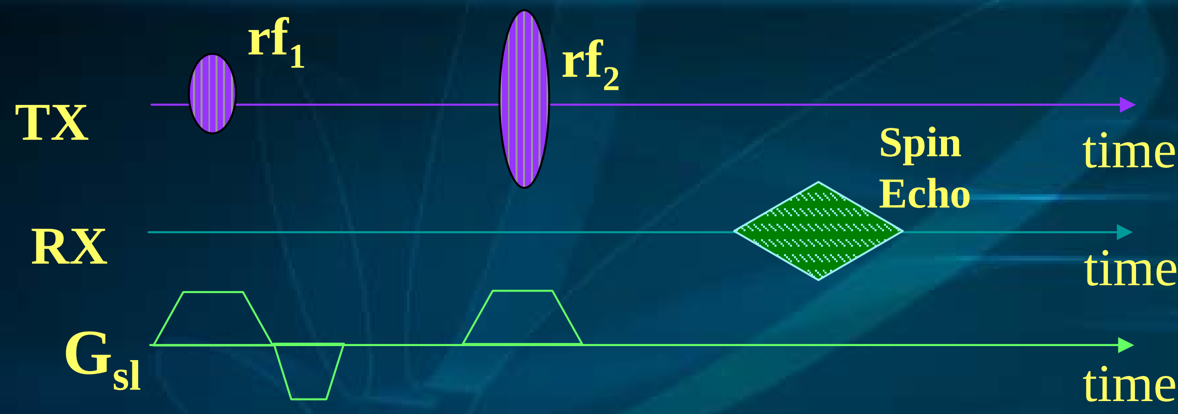
$$slice\ thickness = \frac{2\pi}{\gamma} \cdot \frac{BW}{G_{ss}}$$

FT Approximation

A *sinc function* ($\frac{\sin x}{x}$) envelope on the r.f. pulse produces a nearly square excitation profile of the phantom…..

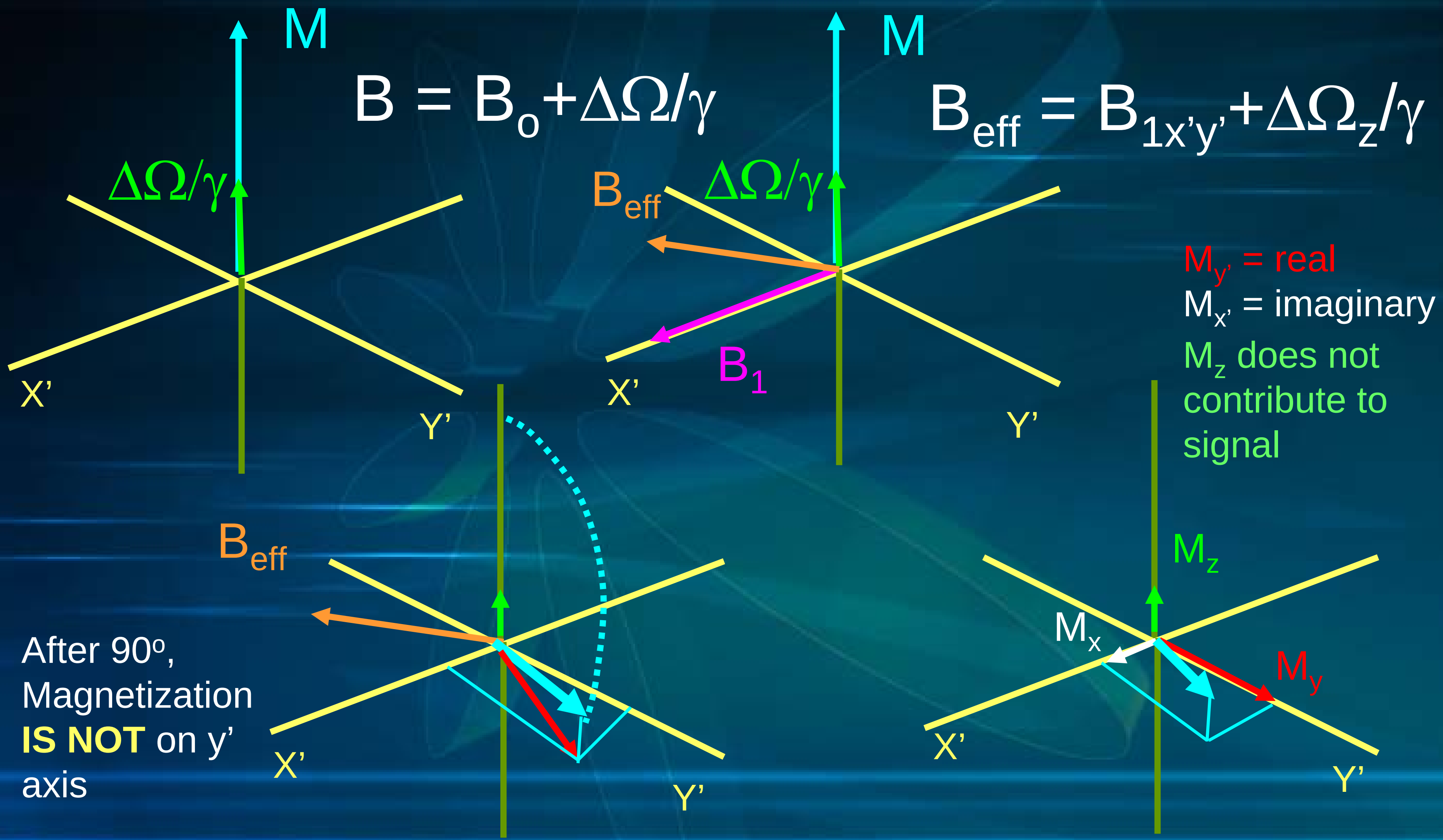


Slice Selection



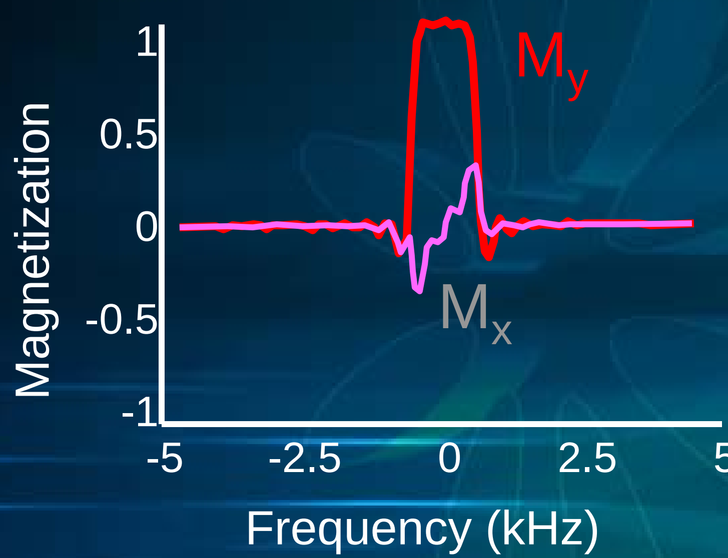
- If constant gradient field is on during the rf pulse:
  - Larmor frequency of spins varies with position
  - The flip angle depends on the local Larmor frequency <u>and</u> the frequency content of the RF field pulse
  - the RF pulse can be “crafted” to contain frequencies in only a specified range

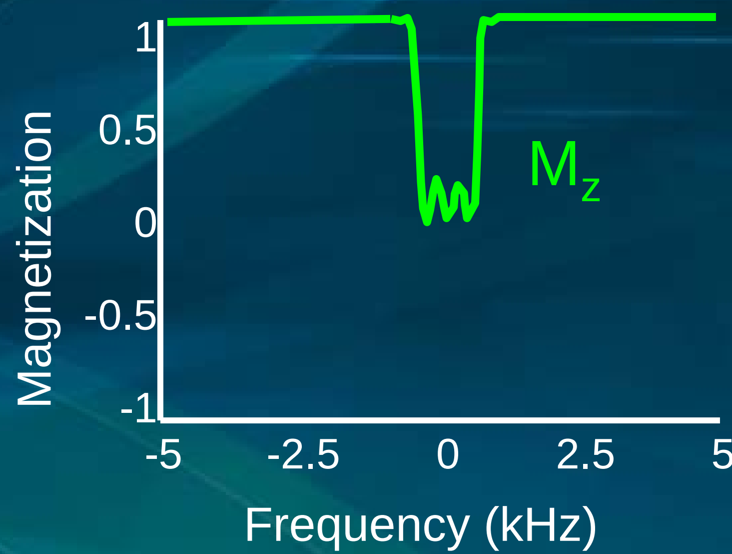
Resonant Frequency Offset



After 90°, Magnetization **IS NOT** on y' axis

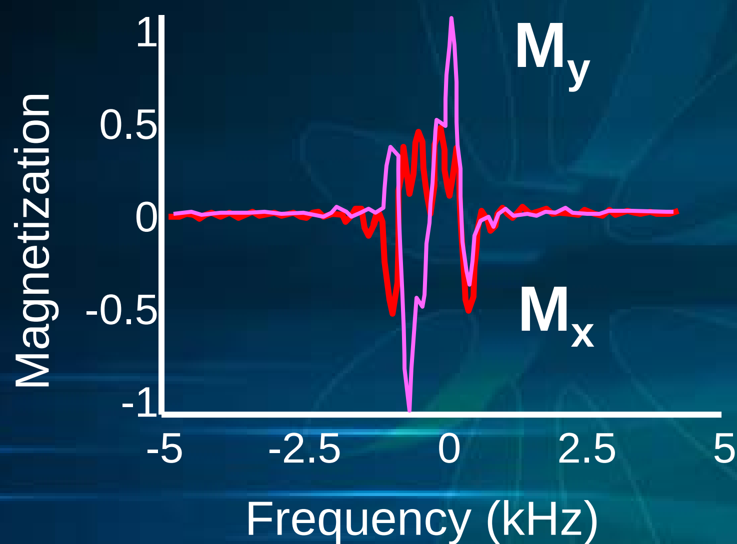
90º Sinc Pulse Profile

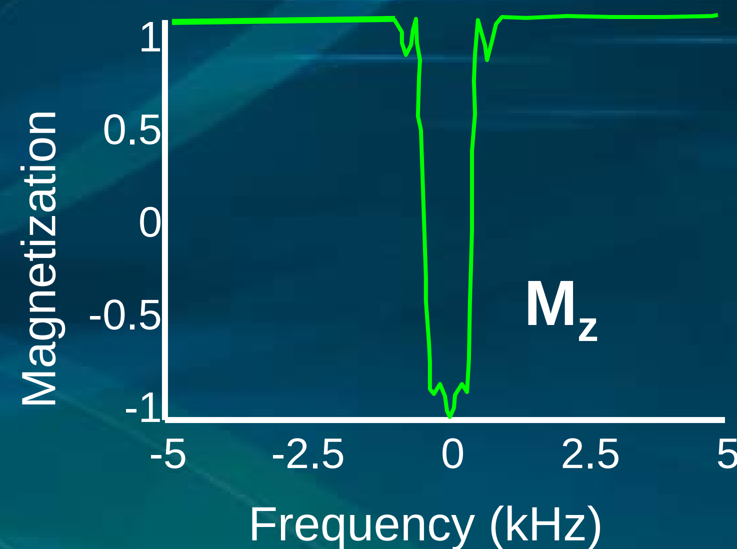




2 ms, 5-lobe, chemical shift refocused

180º Sinc Inversion Pulse





2 ms, 5-lobe, chemical shift refocused

Slice Profile Variations

- Flip Angle varies with location
  - Due to B_1, B_0 field nonuniformities
- Non-linearity of Excitation (Bloch Eqns)
  - FT approximation invalid for big flip angles
  - Bloch simulator software

http://www-mrsrl.stanford.edu/~brian/mritools.html *

- T1-weighting of excitation profile

*Brain Hargreaves

Bloch Equations

- A set of simultaneous differential equations that describe the behavior of the magnetization under any conditions.

$$\frac{dM_z(t)}{dt} = \gamma[M_x(t)B_y(t) - M_yB_x(t)] - \frac{M_z(t) - M_0}{\mathrm{T}_1}$$

Magnetization along the z-axis

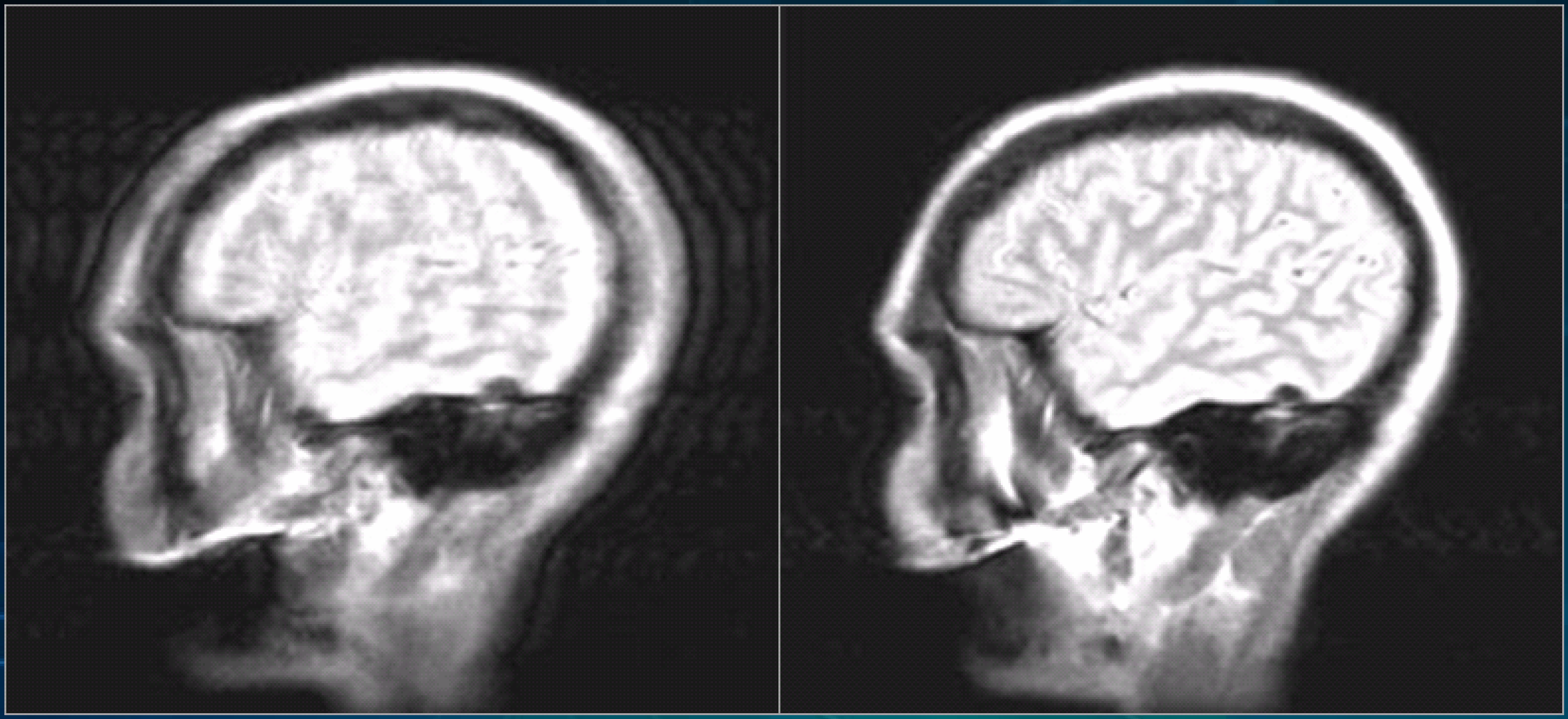
$$\frac{dM_x(t)}{dt} = \gamma[M_y(t)B_z(t) - M_zB_y(t)] - \frac{M_x(t)}{\mathrm{T}_2}$$

Magnetization along the x-axis

$$\frac{dM_y(t)}{dt} = \gamma[M_z(t)B_x(t) - M_xB_z(t)] - \frac{M_y(t)}{\mathrm{T}_2}$$

Magnetization along the y-axis

Poor RF Pulse Calibration



Miscalibration of FSE 180º RF pulses (left image) is corrected (right image)

B1 Field Mapping - Purpose

a. **Needed for accurate measurement of many NMR parameters, i.e. relaxation times**

b. **Enables estimation of systematic errors in parameter measurement**

c. **Enable correction of spatial sensitivity variation using reciprocity**

B1 Field Mapping - Methods

a. **One-pulse read $M_{x,y}$**

Venkatsen et al. Magn Reson Med 1998; 40:592

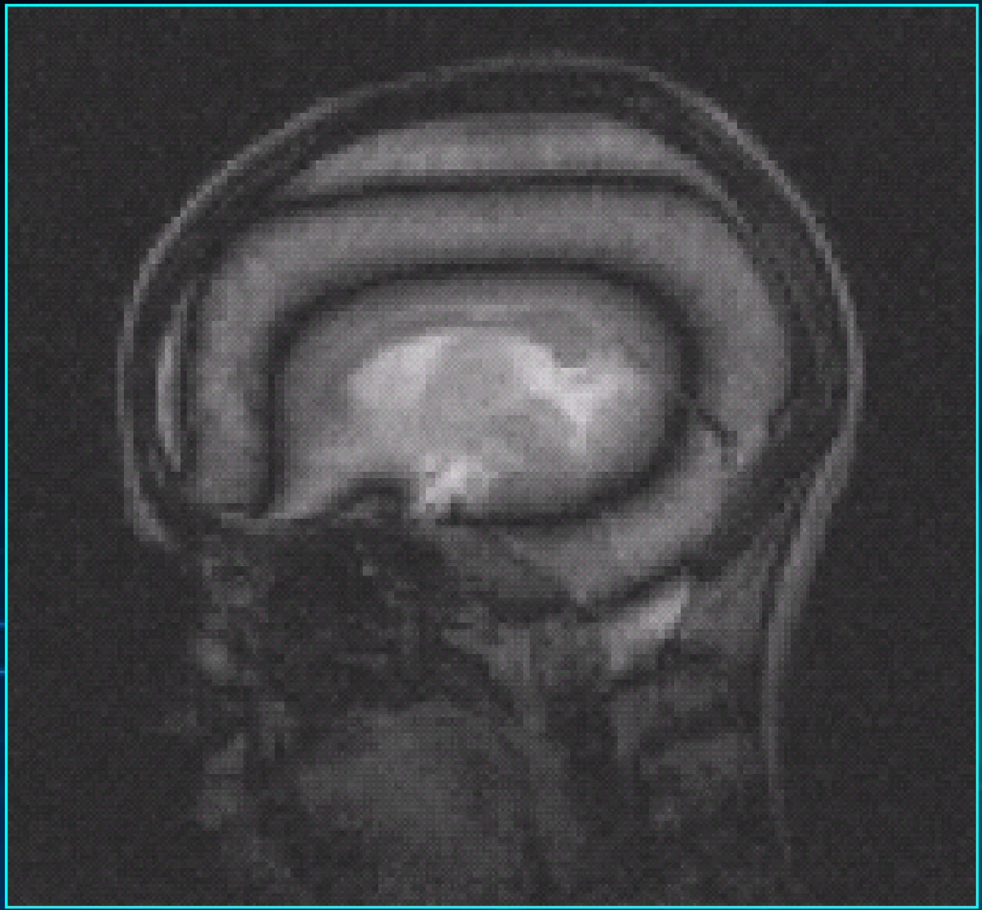
b. **Spin Echo (both pulses altered)**

Barker et al. BJR 1998; 71: 59-67

c. **One-pulse read M_z**

Vaughn et al. Mgn Reson Med 2001; 46: 24

B1 Field Mapping



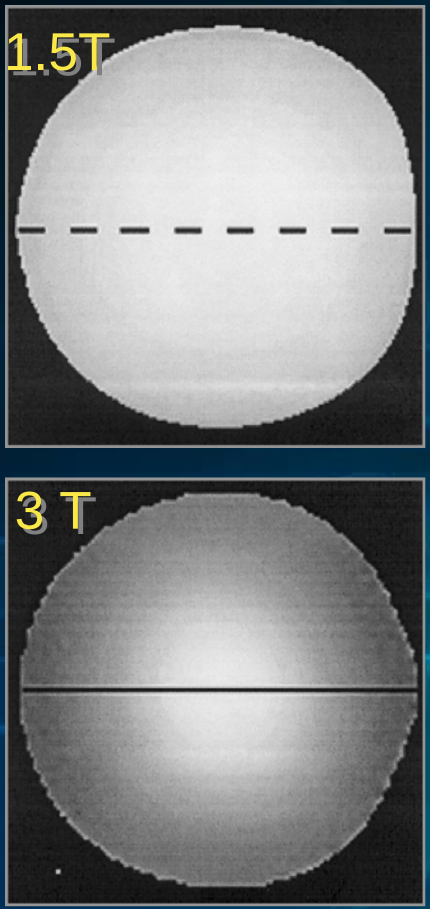
- One-pulse $M_{x,y}$ method
- Hard 1800º pulse preceding 2D field echo sequence
- Bright center is maximum B_1
- Ring pattern occurs at every 5% change in B_1-field

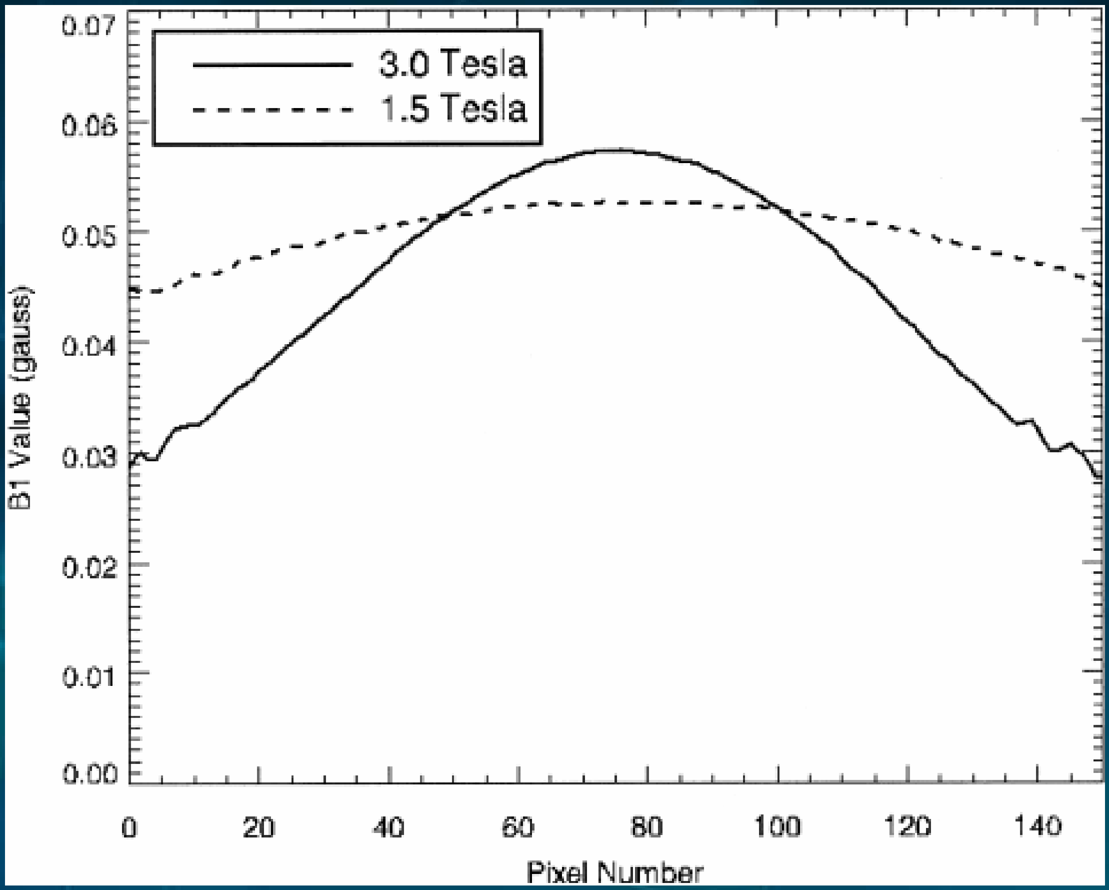
Deichmann R et al. Magn Reson Med 2002; 47: 398

Dielectric Resonance Effect

- Fields inside a sample of diameter, d, can be resonant when $d = n\lambda$
- This can make the B1 field at the center of the sample larger than at the edges
- This effect is dampened as the conductivity of the sample increases

Image Uniformity at 3 Tesla

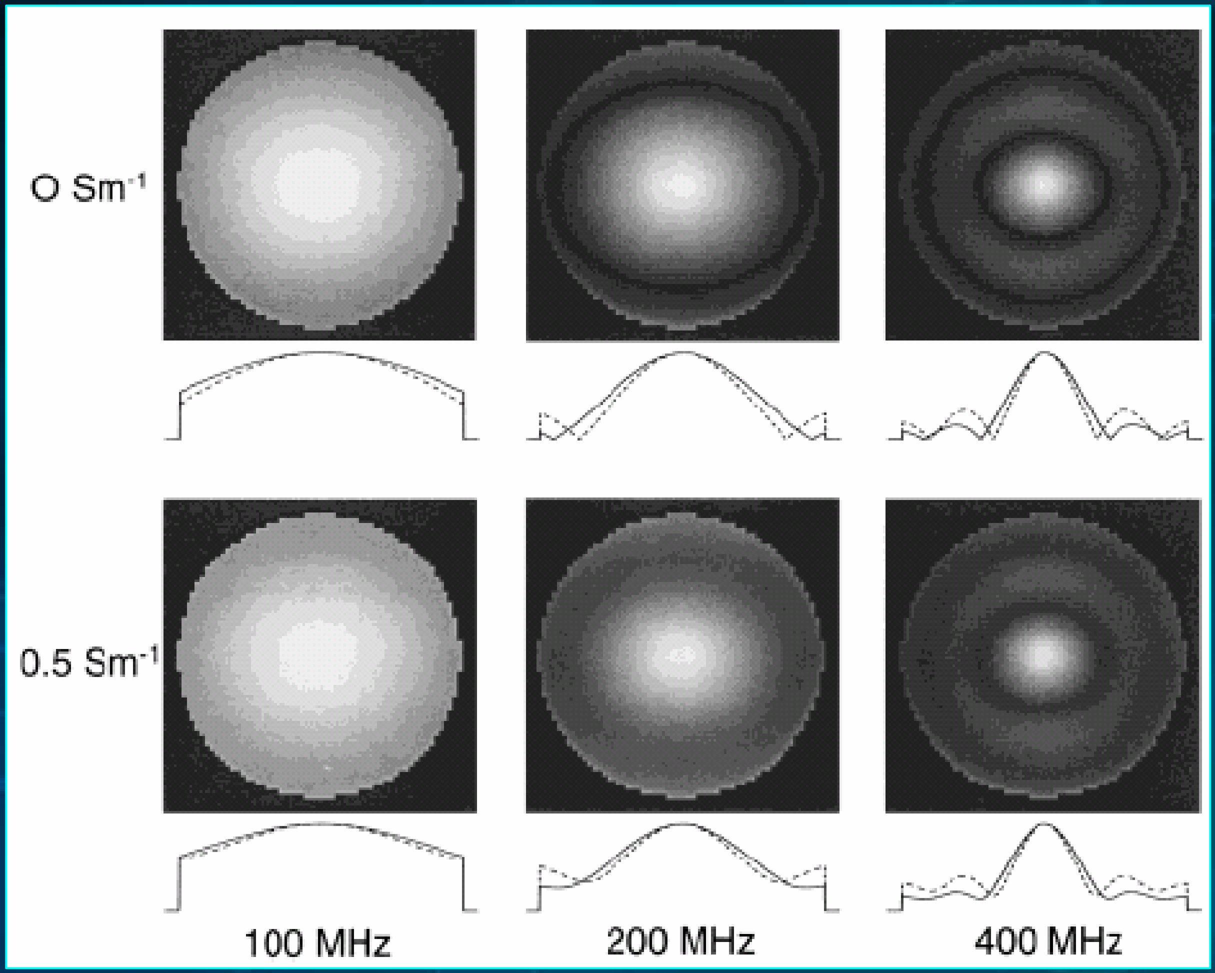




- B1 field maps in a saline phantom (18 cm diameter)

RL Greenman et al. JMRI 2003, 17(6): 648-655

Dielectric Resonance



Hoult DI, J Magn Reson Imag, 2000; 12:46-67

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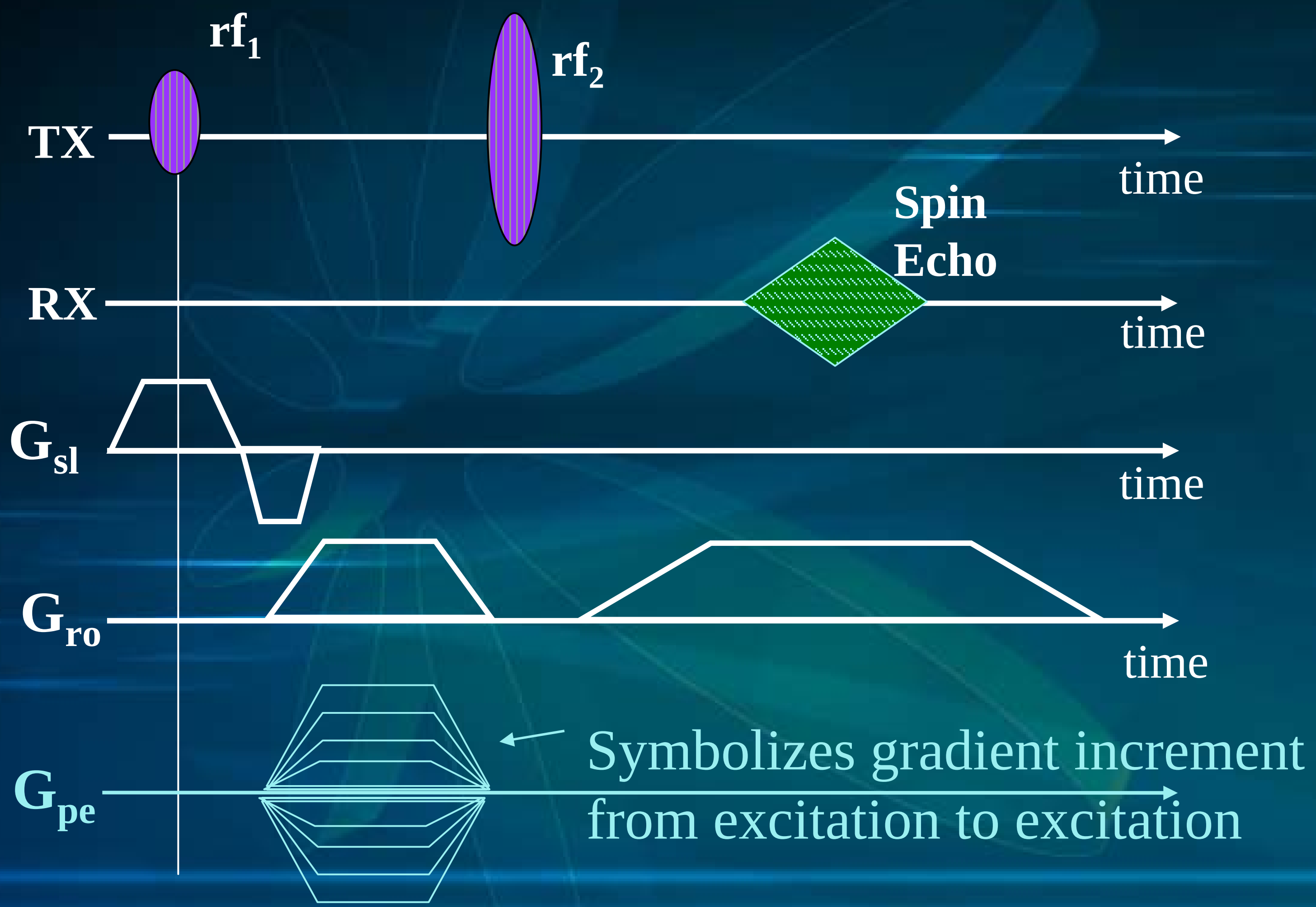
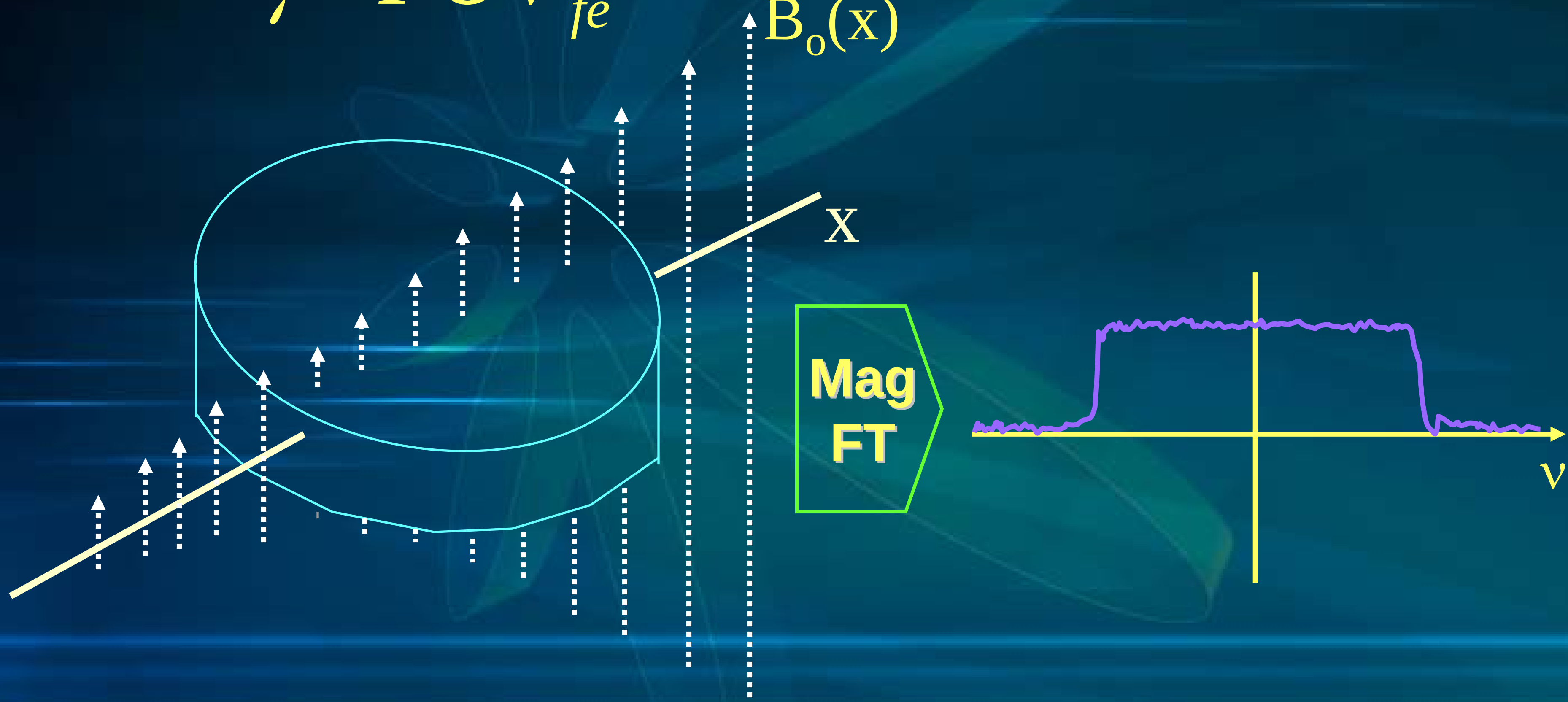
Spin-Echo Sequence

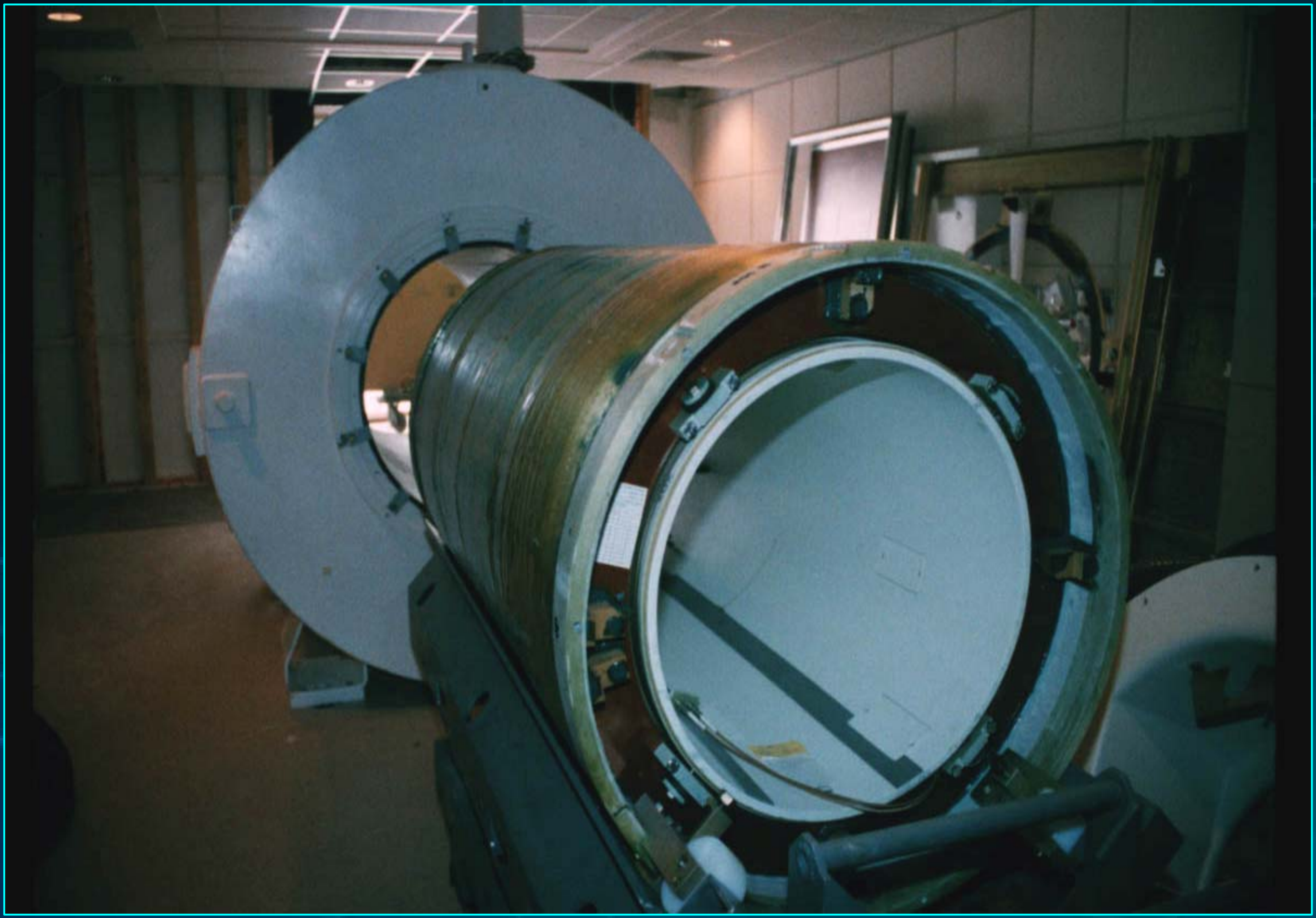


One Dimensional FT

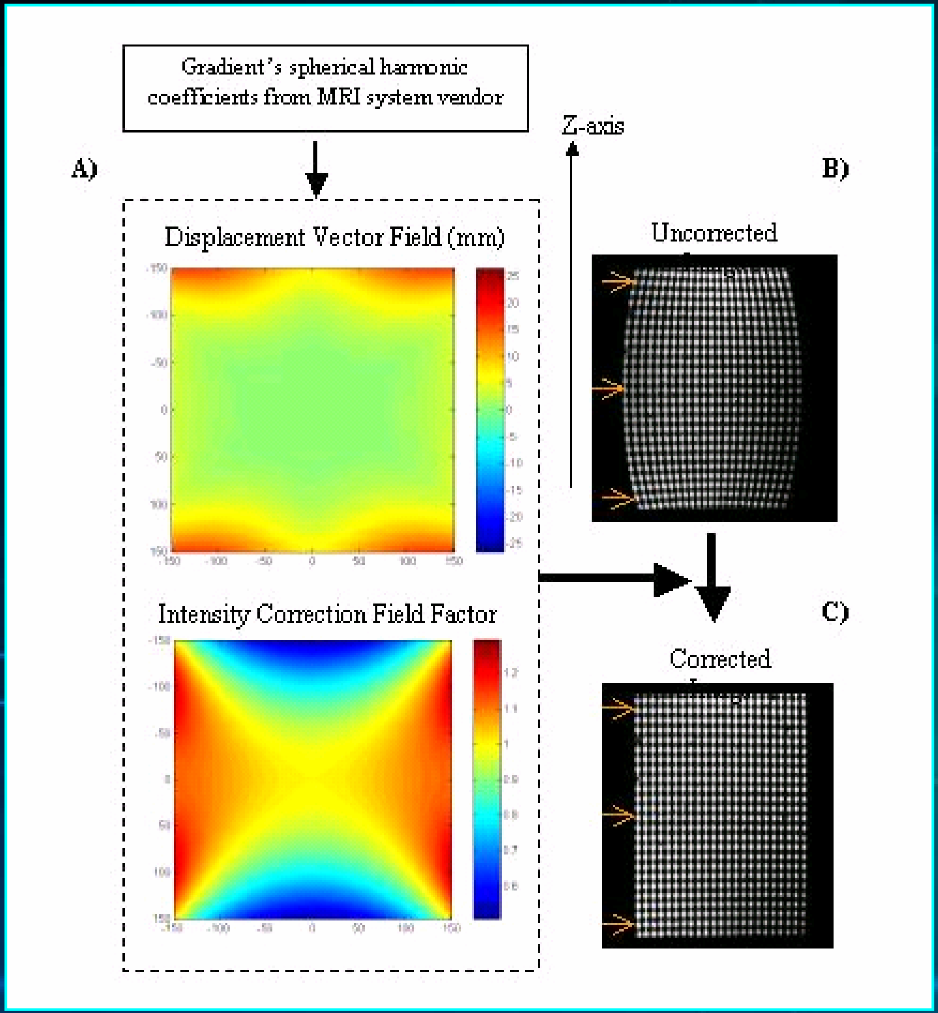
$$G_{fe} = \frac{2\pi}{\gamma}\frac{BW}{FOV_{fe}}$$



Gradient Subsystems



MRI Gradient Fields



- Gradient Nonlinearities are often tolerated as part of trade-offs with gradient field strength or coil size
- Manufacturers often apply gradient distortion corrections in order to make images appear to be distortion free
- Influences image quality parameters (SNR, spatial resolution, etc.)

http://www.nbirn.net

MRI Gradient Fields

Sources of Error:

- Gradient Amplitude Calibration
  - –Best about 1% error
- Gradient Non-linearities
- Eddy Currents

Eddy Currents

- Accelerating current in gradient coils (gradient pulse) causes induced currents in nearby metallic structures.
- These currents produce magnetic fields which, in turn, oppose the magnetic fields of the gradient coils

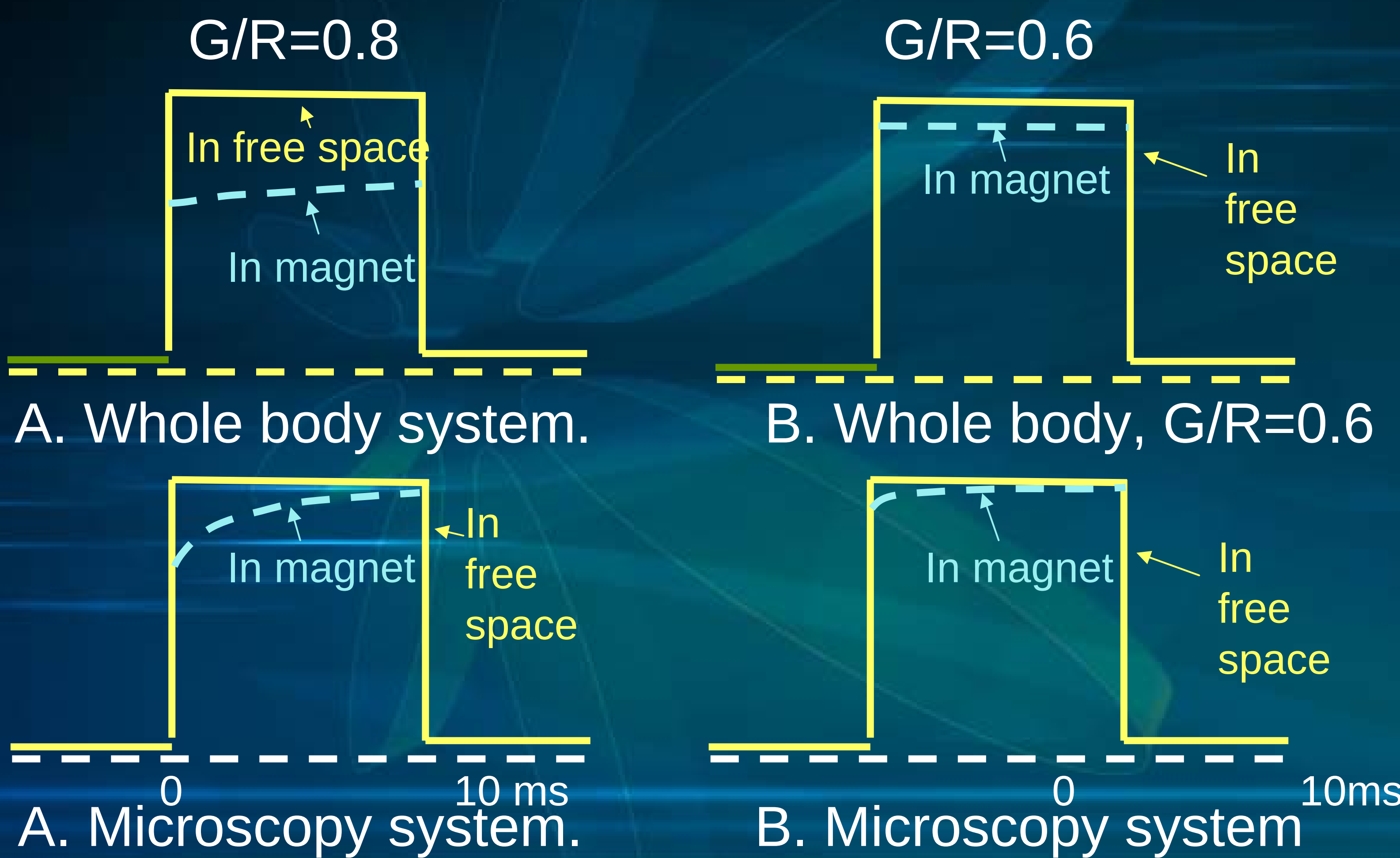
Eddy Currents

- The magnetic field produced by Eddy Currents have two time-dependent components:
  - An offset of the B_o field
  - An additional gradient field

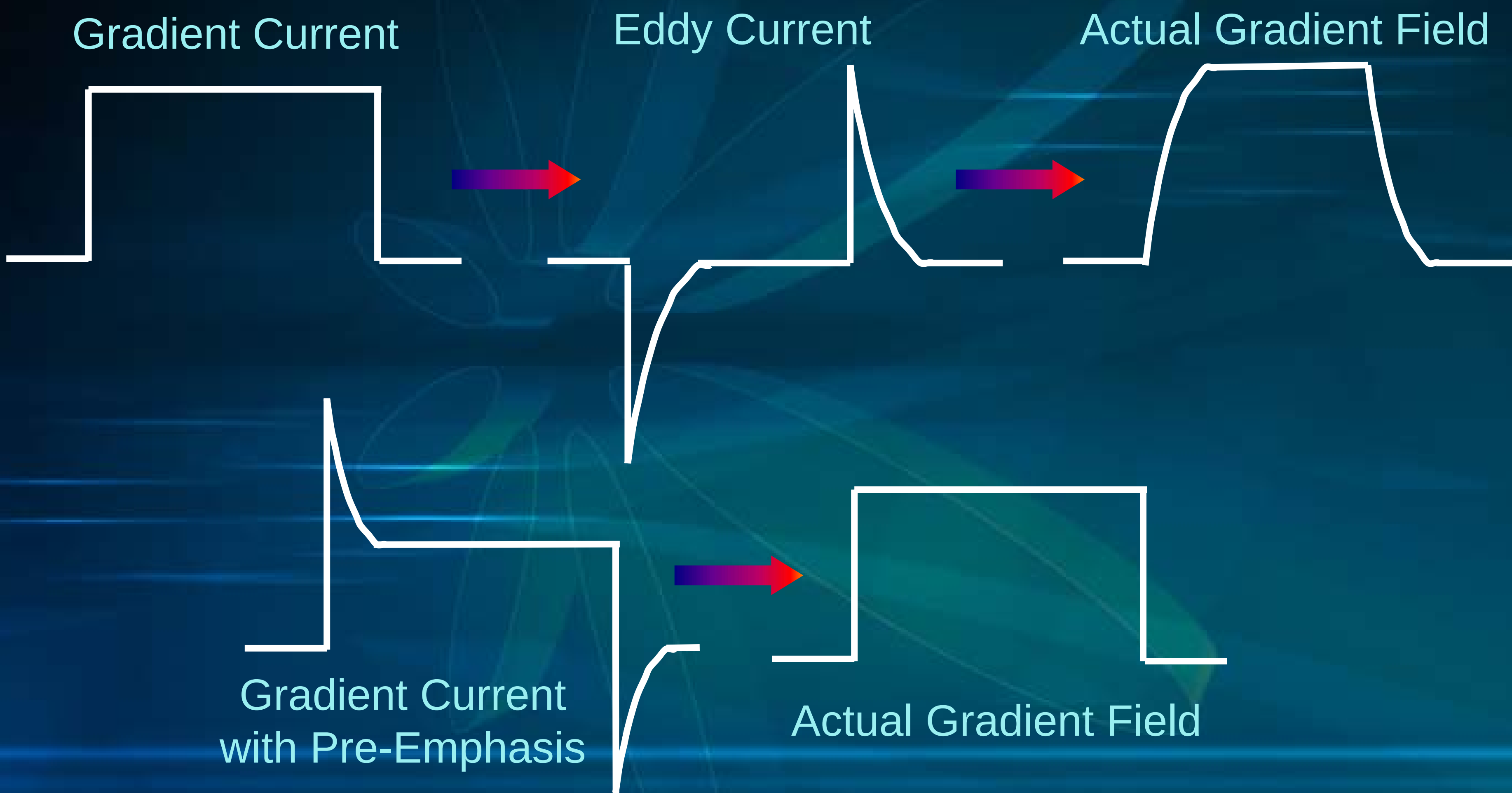
$$B_{ec}(\vec{r},t) = \Delta g(\vec{r},t) + \Delta B_o(t)$$

Gradient Waveforms

Gradient to Magnet Ratio



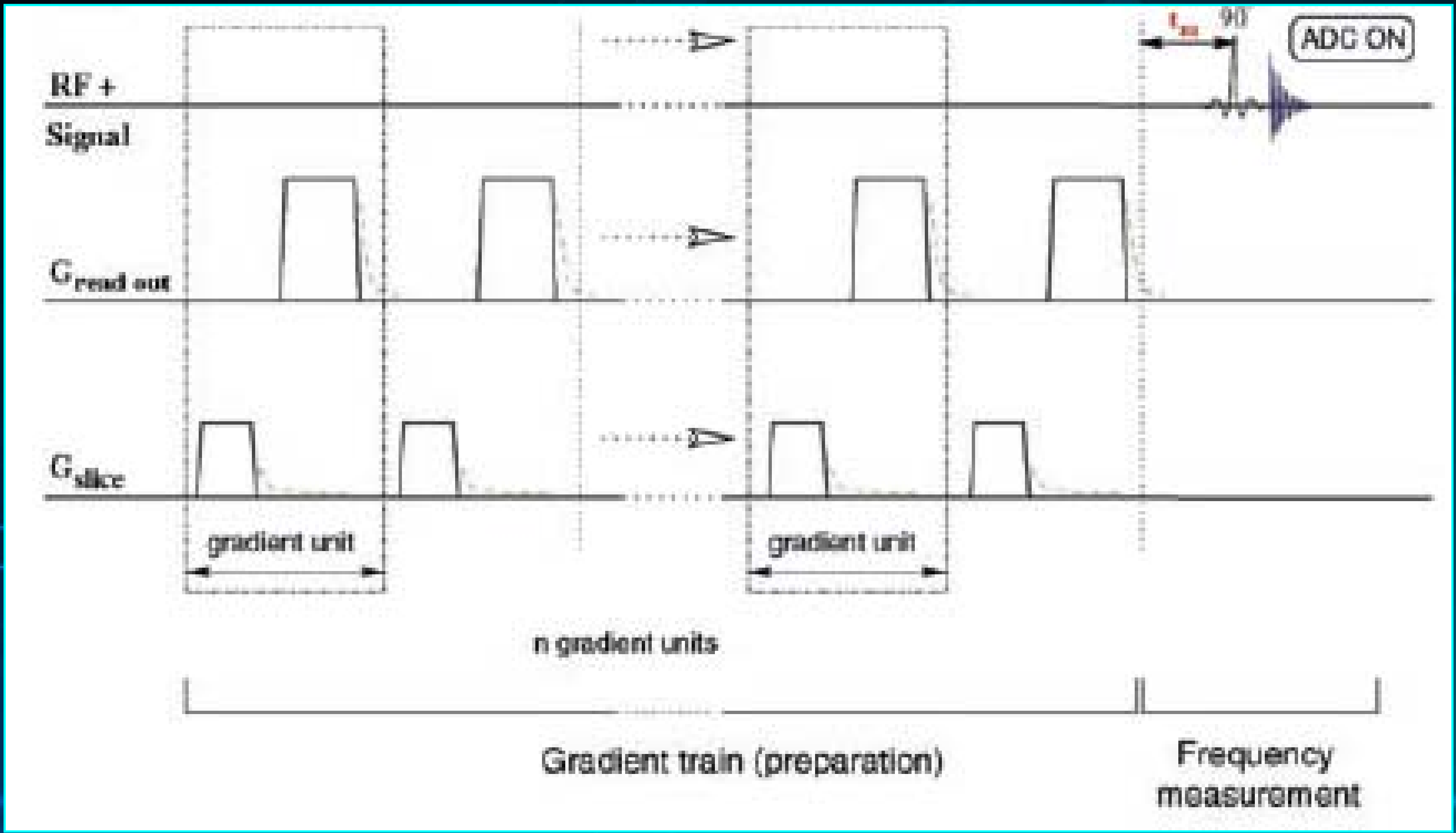
Eddy Current Preemphasis



Actively Screened Gradients

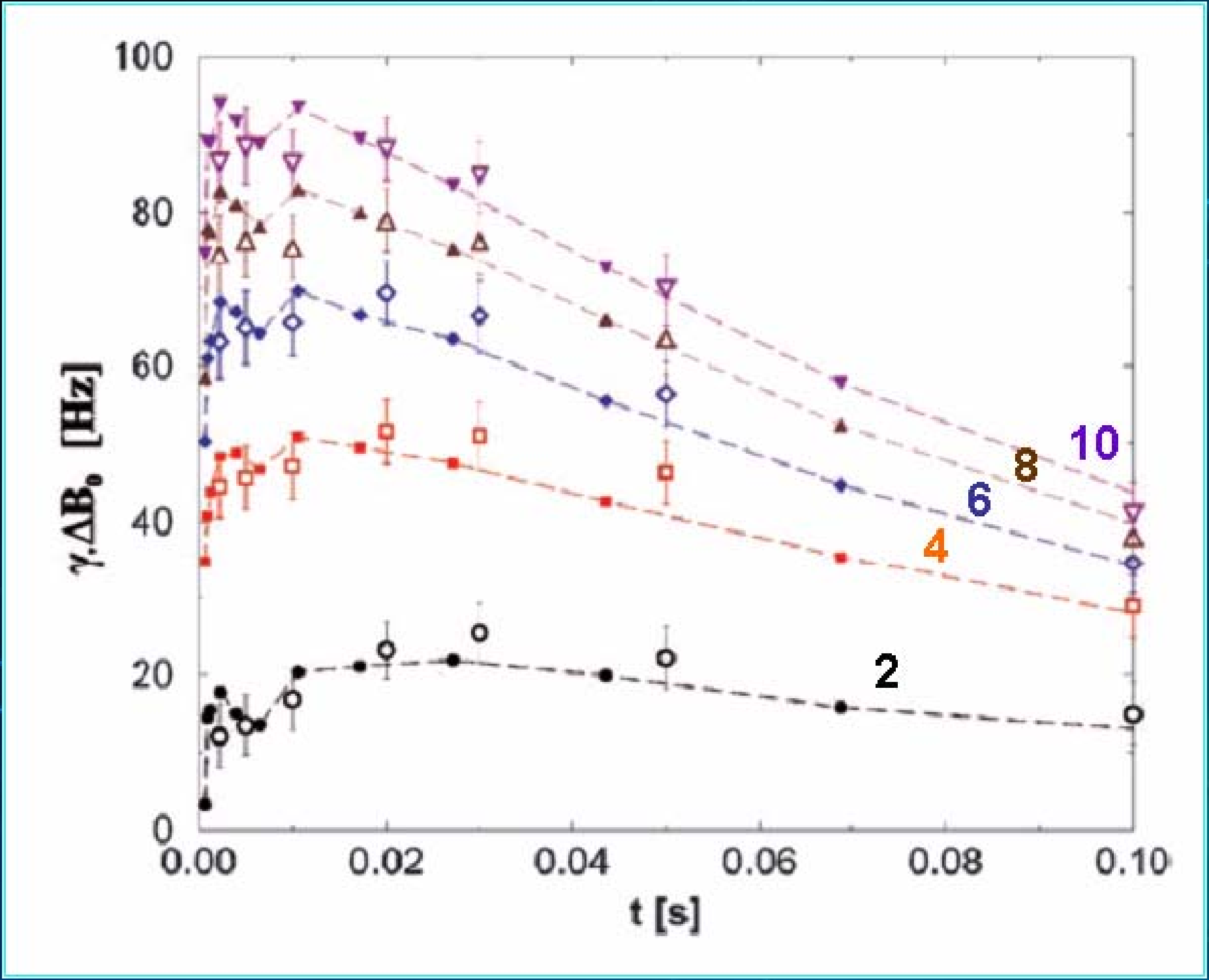
- Reduce gradient field strength outside of gradient coil former
  - Current in shield is opposite polarity
- Reduces gradient field in imaging volume also
  - Improves magnet homogeneity
- Each gradient coil is associated with a screen coil
  - Twice as many amplifiers required

Measuring Eddy Currents



De Deene Y et a. Phys Med Biol 2000; 45:1807-1823

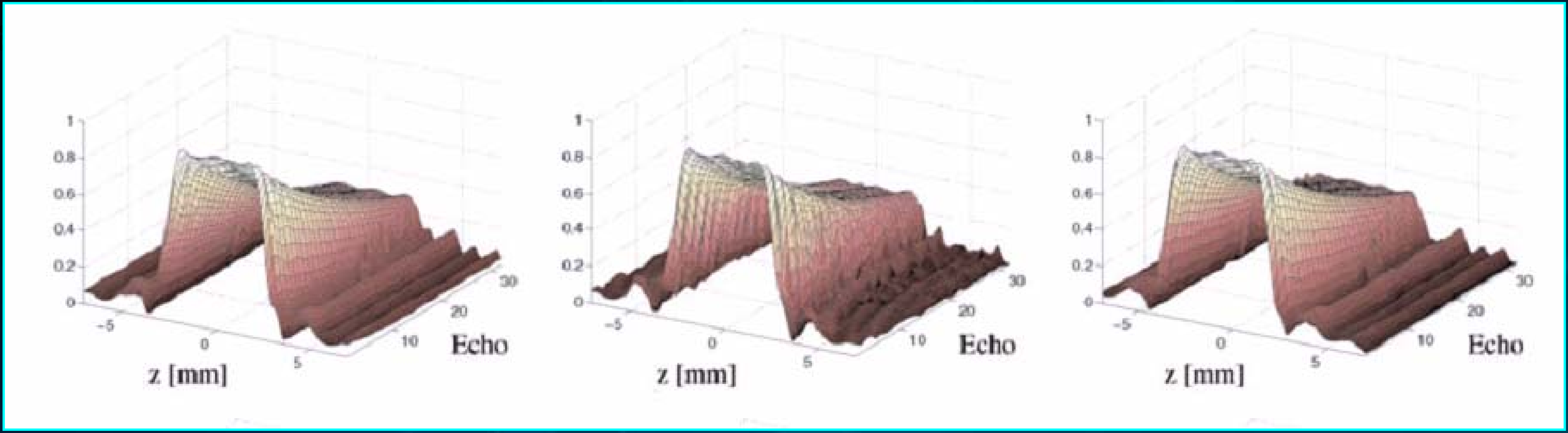
Eddy Currents



Time course of eddy current field offset following different numbers of gradient pulse units

De Deene Y et a. Phys Med Biol 2000; 45:1807-1823

Eddy Current Effects on Slice



Ideal – no eddy currents

worst case

pre-gradient train

RF pulse profiles used for T2-relaxometry a) Ideal profile calculated from Bloch equations b) profile showing the influence of eddy currents; c) pre-gradient pulse train establishes steady-state which regularizes the RF pulse profile

De Deene Y et a. Phys Med Biol 2000; 45:1807-1823

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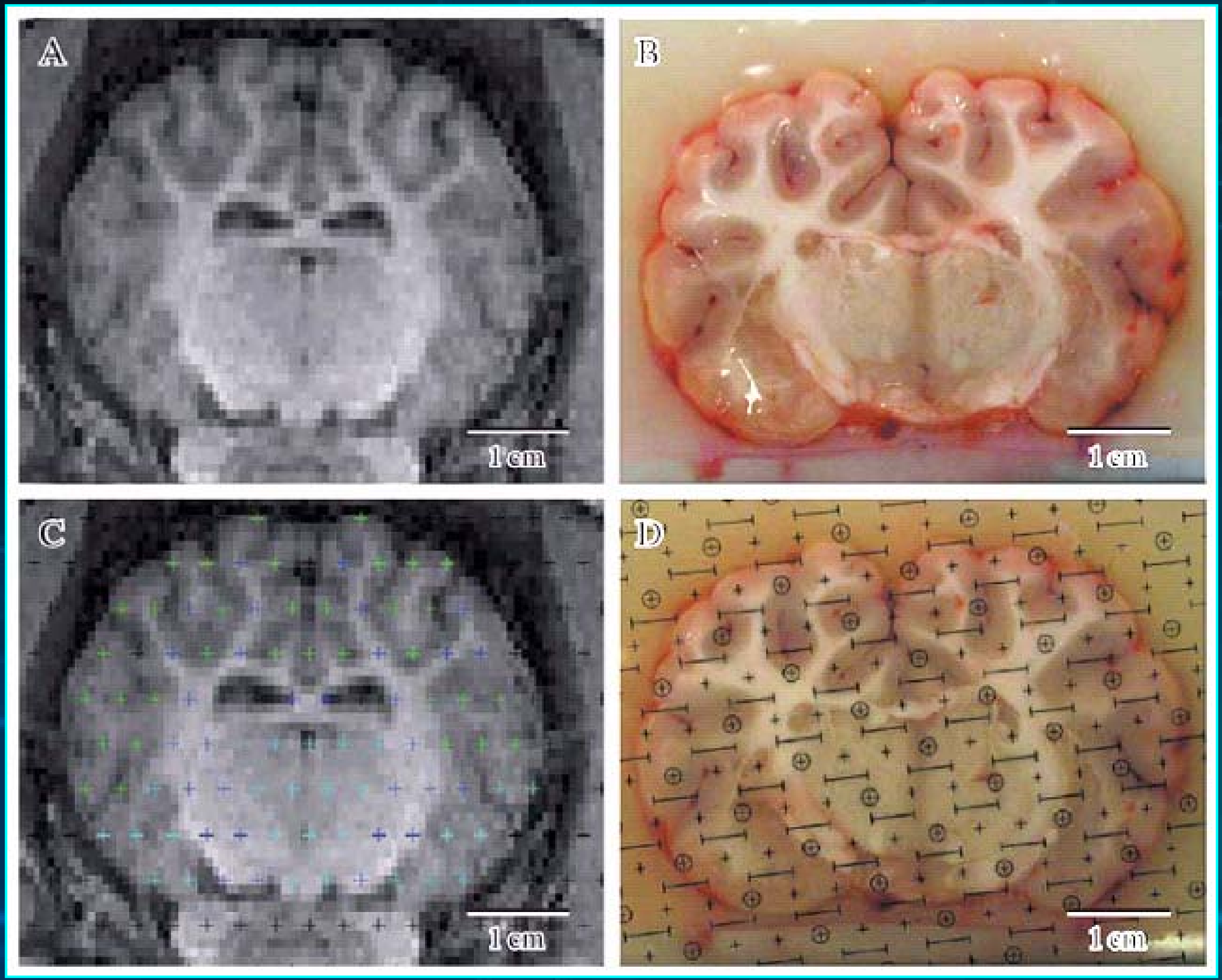
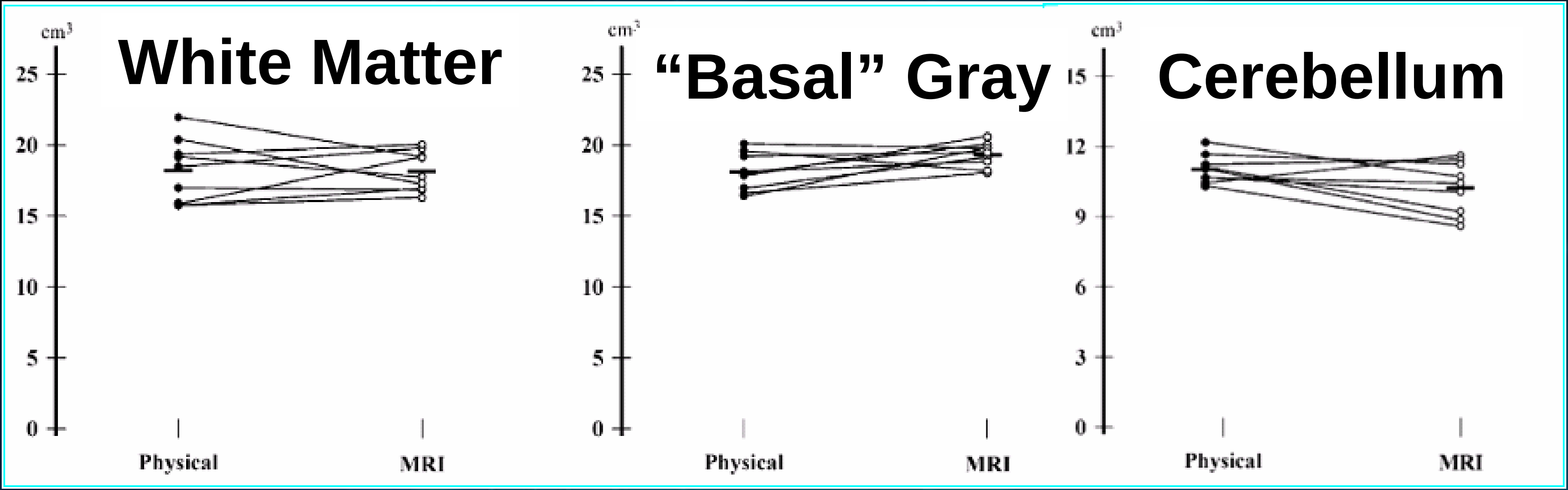
Brain Volume Measurement



Jelsing J NeuroImage 2005; 26: 57-65

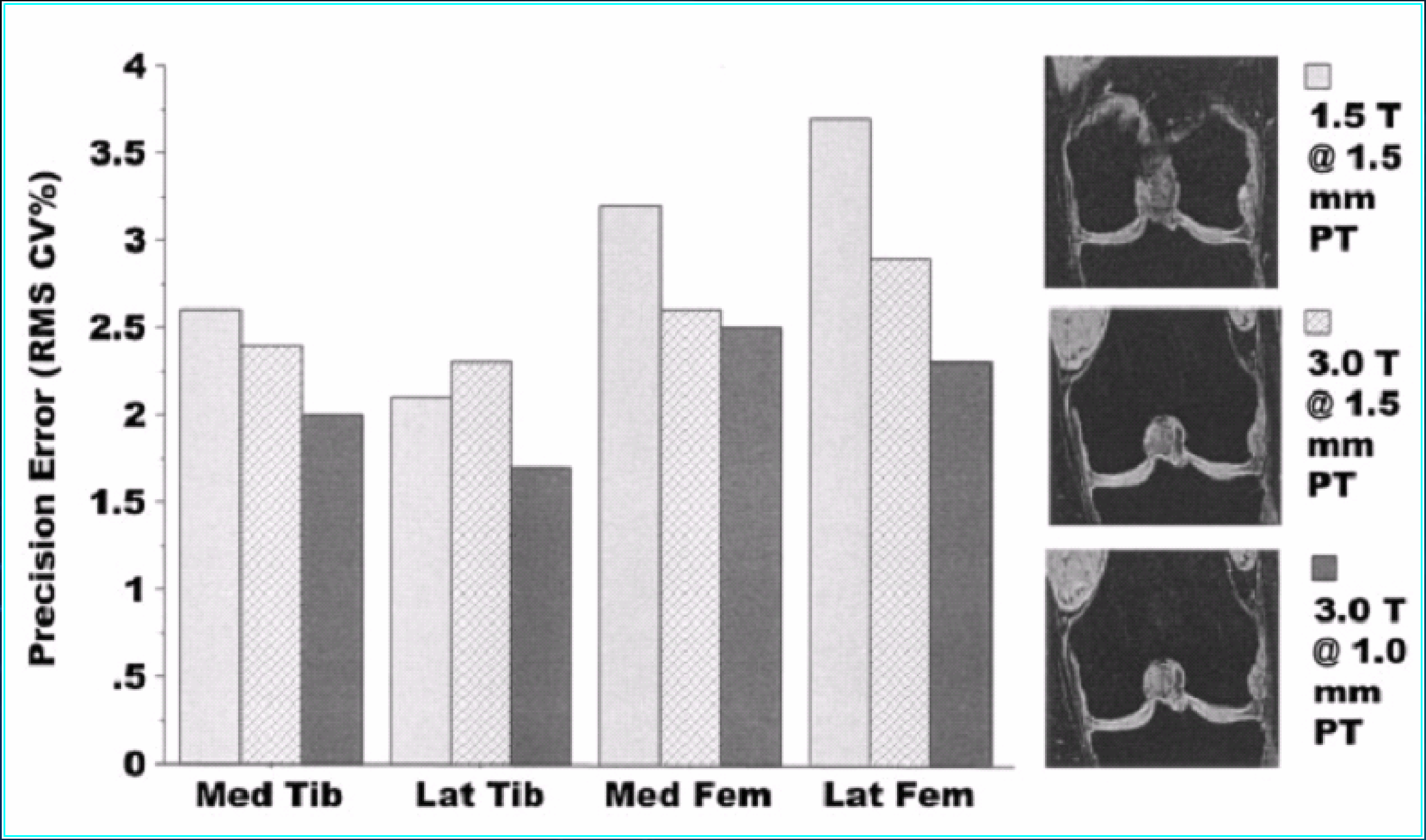
Brain Volume Relationships



- High agreement between qMRI volumetry and physical sections
- qMRI volumetry is susceptible to high inter-observer variability
- Problems greatest in those regions where tissue margins are poorly defined

Jelsing J NeuroImage 2005; 26: 57-65

Cartilage MRI



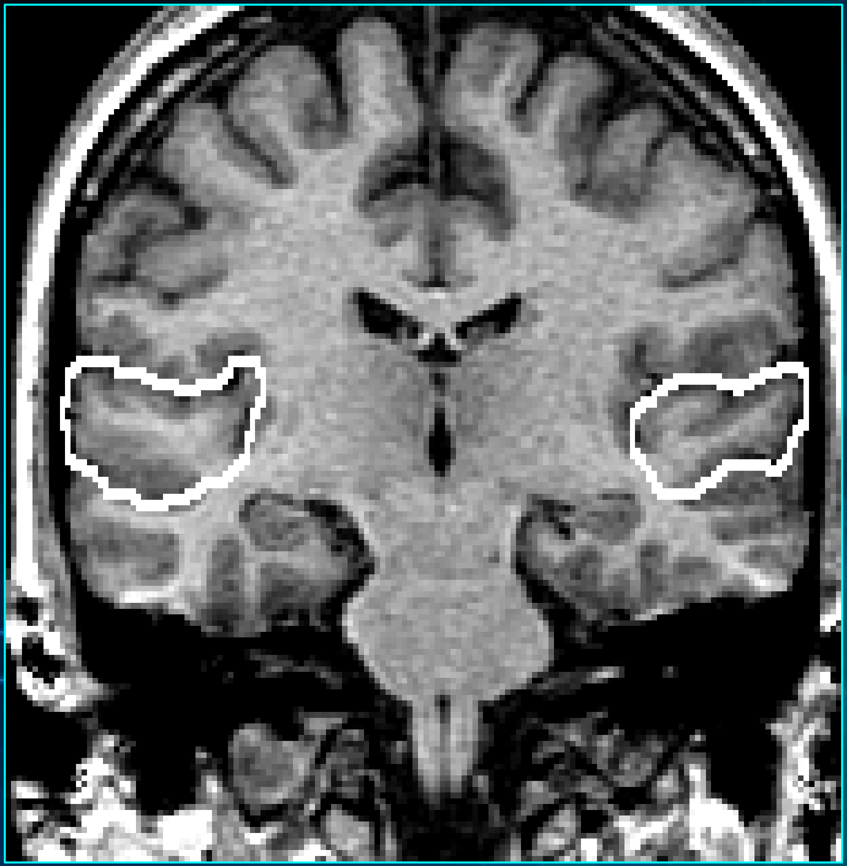
Accuracy of 3T high and tends to be more reproducible than 1.5 T

Eckstein F et al. Arth & Rheum 2005; 52: 3132-3136

MRI in Psychiatry

- MRI is a safe and non-invasive technique to look into the to measure the human brain's normal anatomy *in vivo*
- Used in the investigation of the pathophysiology of mood disorders:
  - major depression
  - bipolar disorder
- Potential for the evaluation of psychiatric therapeutic responses

MRI Volume Measurements



Superior temporal gyrus tracing

Pituitary gland tracing

Volumes in Bipolar Disease

	Bipolar	Healthy Controls	F	p
Total left STG volume (cm^3)	12.5±1.5	13.6±2.5	4.45	**0.043**
White matter left STG volume (cm^3)	2.79±0.56	3.12±0.73	4.23	**0.048**
Gray matter left STG volume (cm^3)	9.7±1.3	10.5±2.1	2.36	0.134
Total right STG volume (cm^3)	14.9±2.3	15.3±1.9	0.57	0.454
White matter right STG volume (cm^3)	4.60±0.95	4.99±0.98	4.85	**0.035**
Gray matter right STG volume (cm^3)	10.3±1.8	10.3±1.5	0.01	0.930

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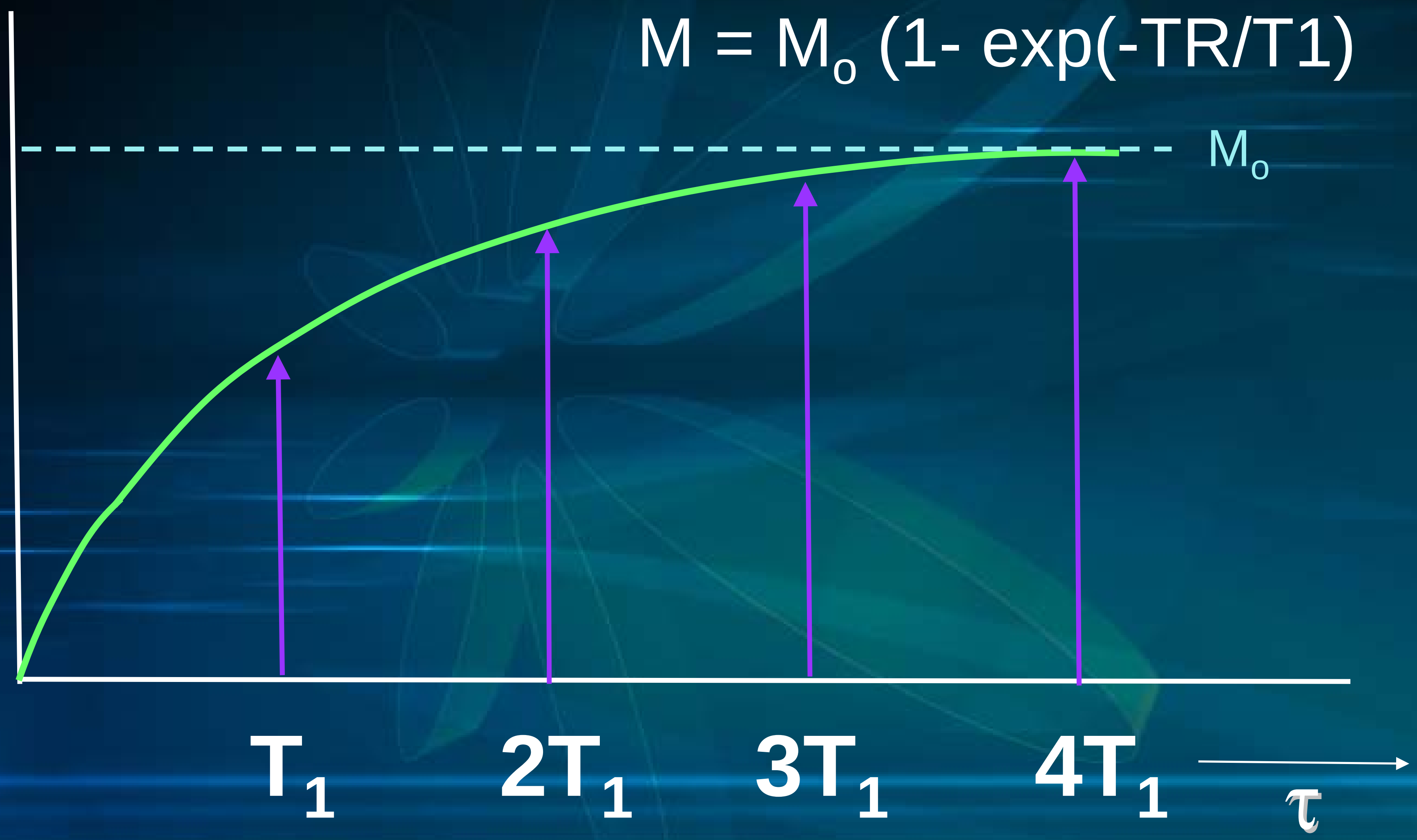
Relaxation Times

- T_1: longitudinal relaxation time defines recovery of potential for next signal ($T_1=1/R_1$)
- T_2: transverse relaxation time defines rate of dephasing of MRI signal due to microscopic processes ($T_2=1/R_2$)
- T_2*: transverse relaxation time with B_o inhomogeneity effects added; defines rate of dephasing of MRI signal due to macroscopic <u>and</u> microscopic processes ($T_2^*=1/R_2^*$)

Longitudinal Relaxation



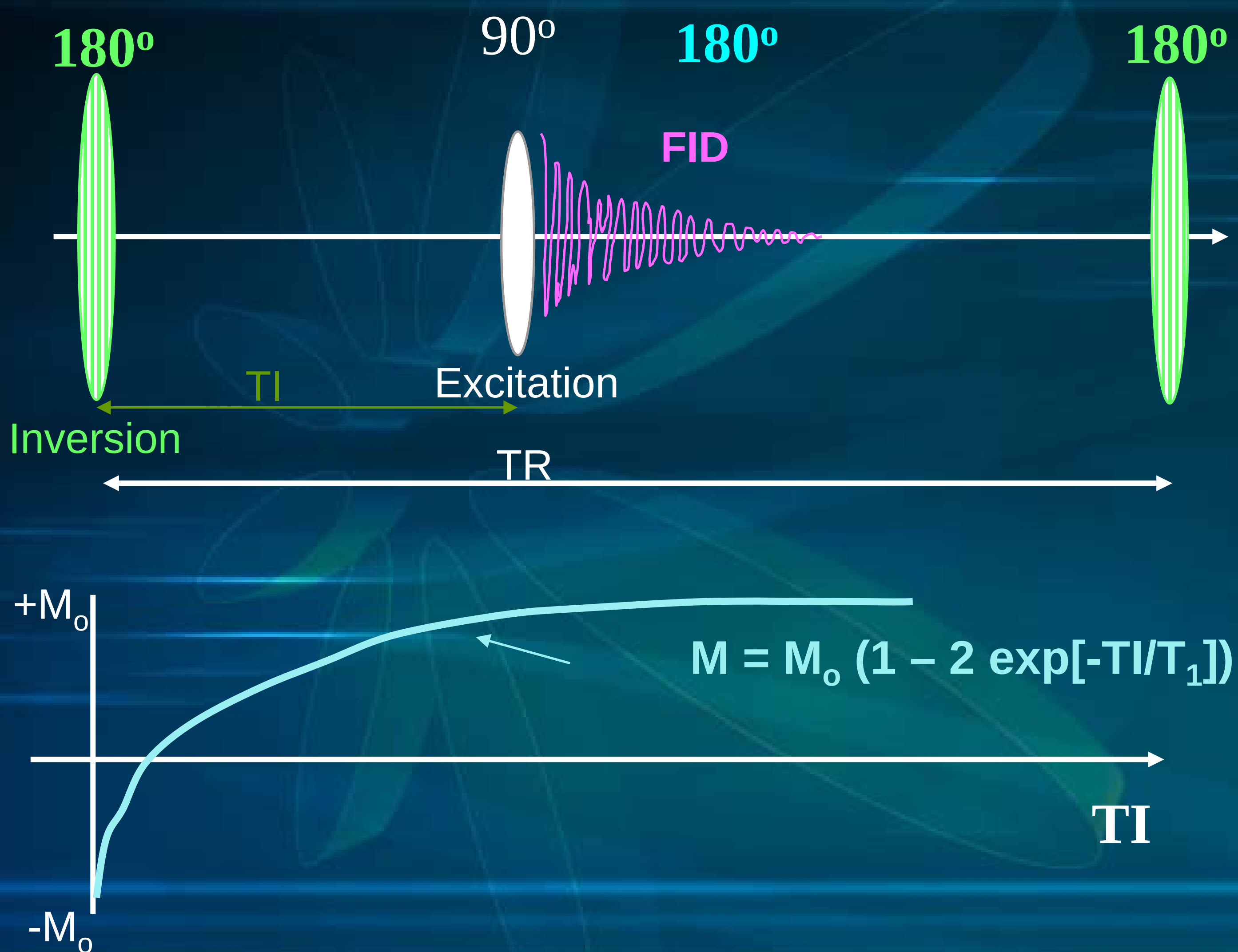
Applications for T1 Images

- Tissue characterization
- Contrast agent uptake studies
- Measurement of Tissue Perfusion
- Measurement of Blood Volume

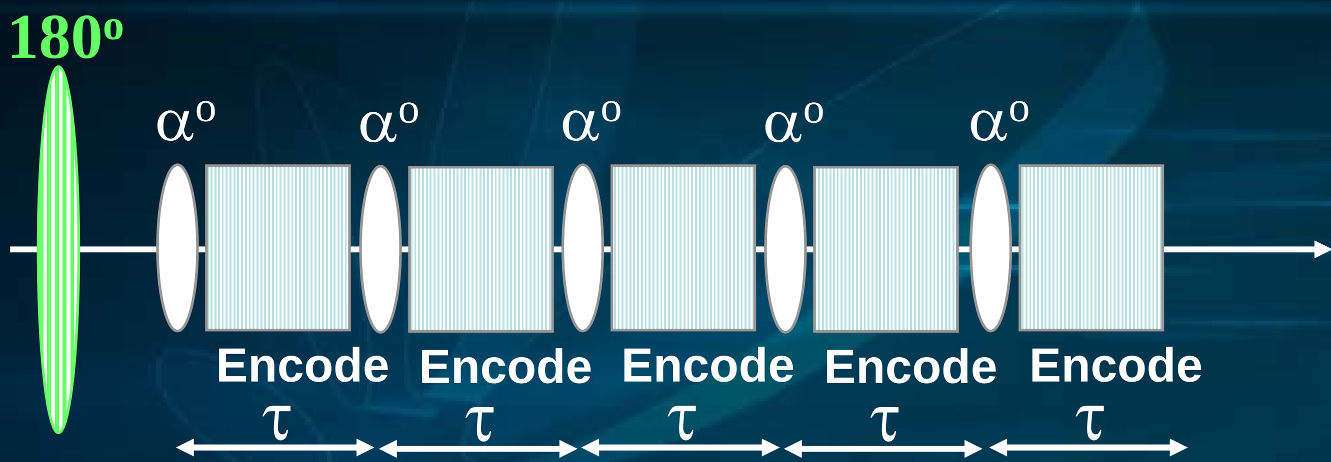
T1 Measurement Sequences

- Inversion Recovery
  - 180º-90º-180º; the gold standard
- Saturation Recovery – 90º-180º
- Stimulated Echo – 90º-90º-90º
- Look-Locker Sequence (see below)

Inversion Recovery



Look-Locker Sequence

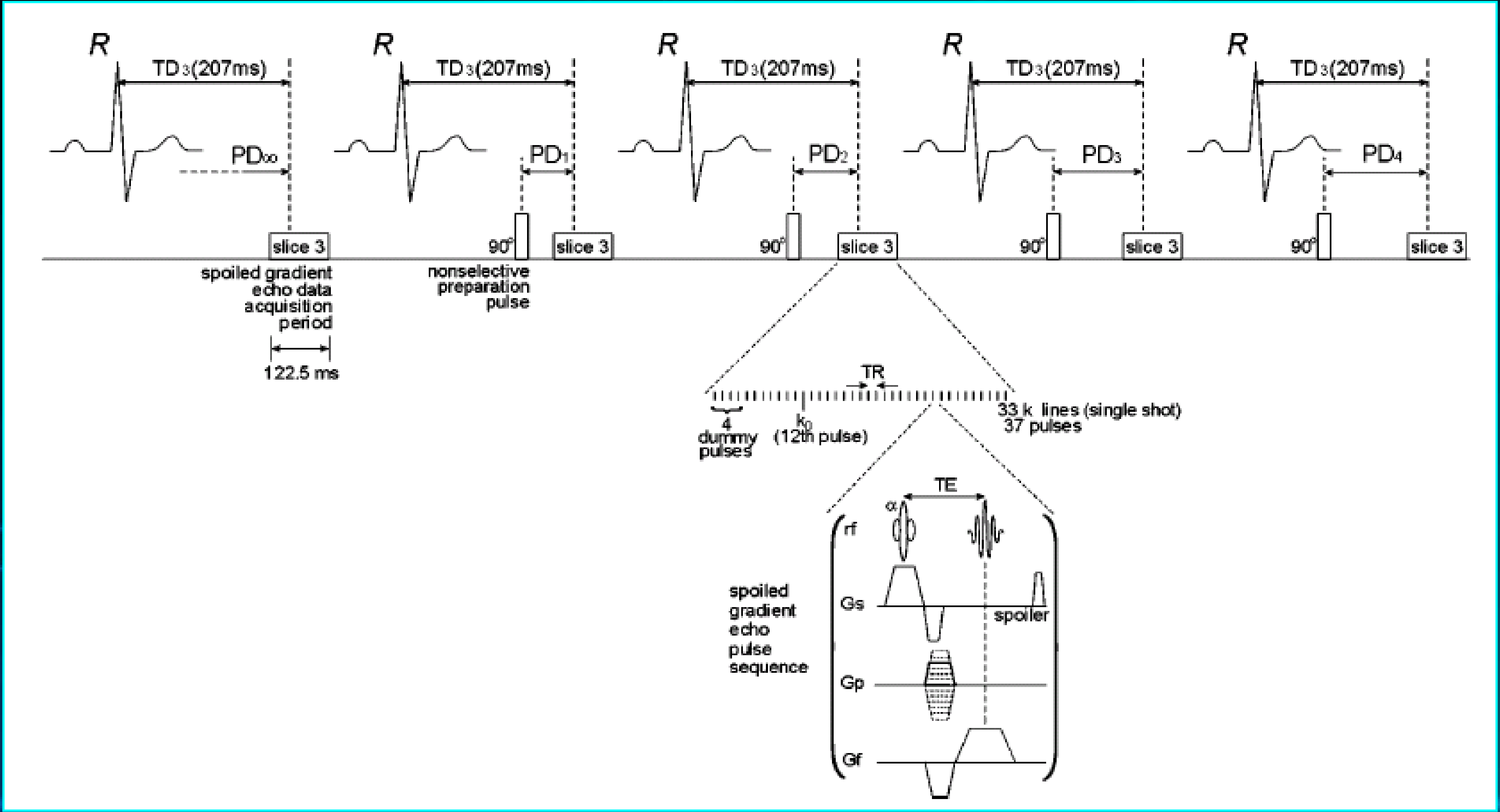


- Very sensitive to RF pulse errors
- Magnetization recovery rate, T_1^* :

$$T_1^* = \frac{\tau}{\left(\tau / T_1\right) - \ln(\cos\alpha)}$$

Look & Locker, Rev Sci Instrum 1970; 41: 250

Saturation Recovery for CMRI



Higgins DM, Med Phys 2005; 32(6):1738-1746

T1 Parametric Maps

(a) T1 Map of tubes of gel doped with Gd-DTPA

(b) T1-weighted image of heart in short axis

(c) T1 parametric map image of heart in (b)

- T1's calculated from short-acquisition period T1 sequence (SAP-T1) with varying delay times

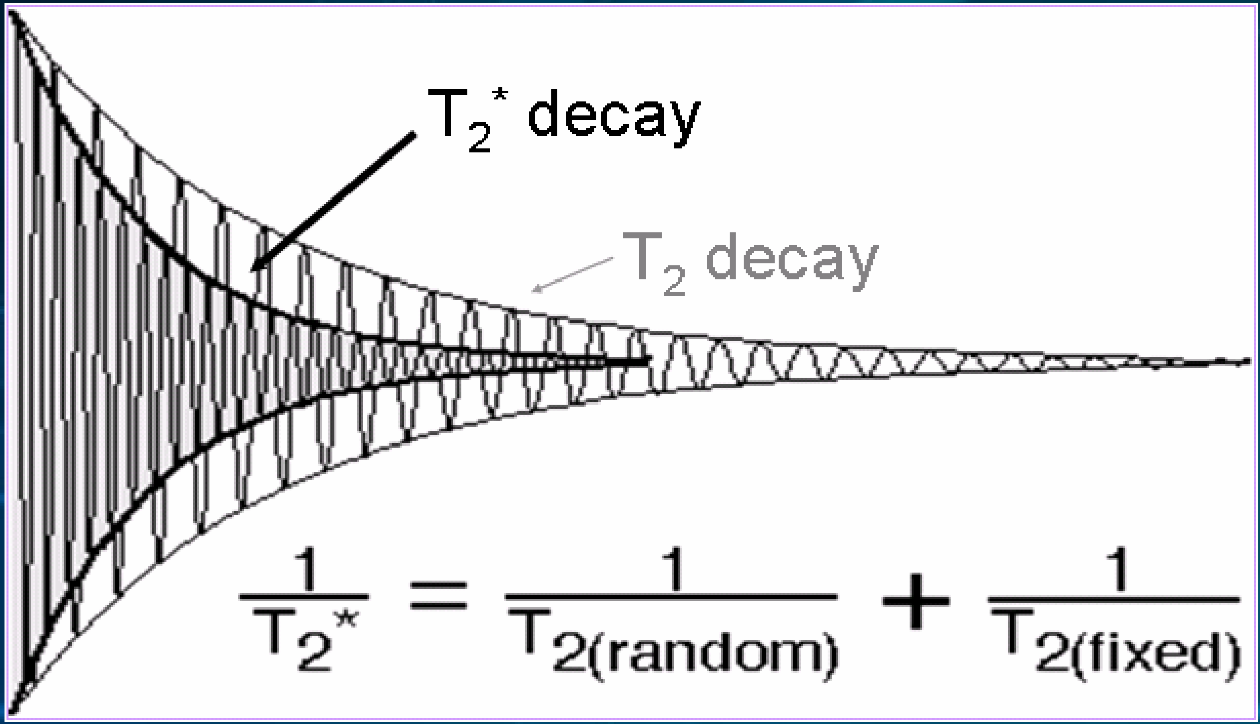
Higgins DM, Med Phys 2005; 32(6):1738-1746

For Accurate & Precise T1

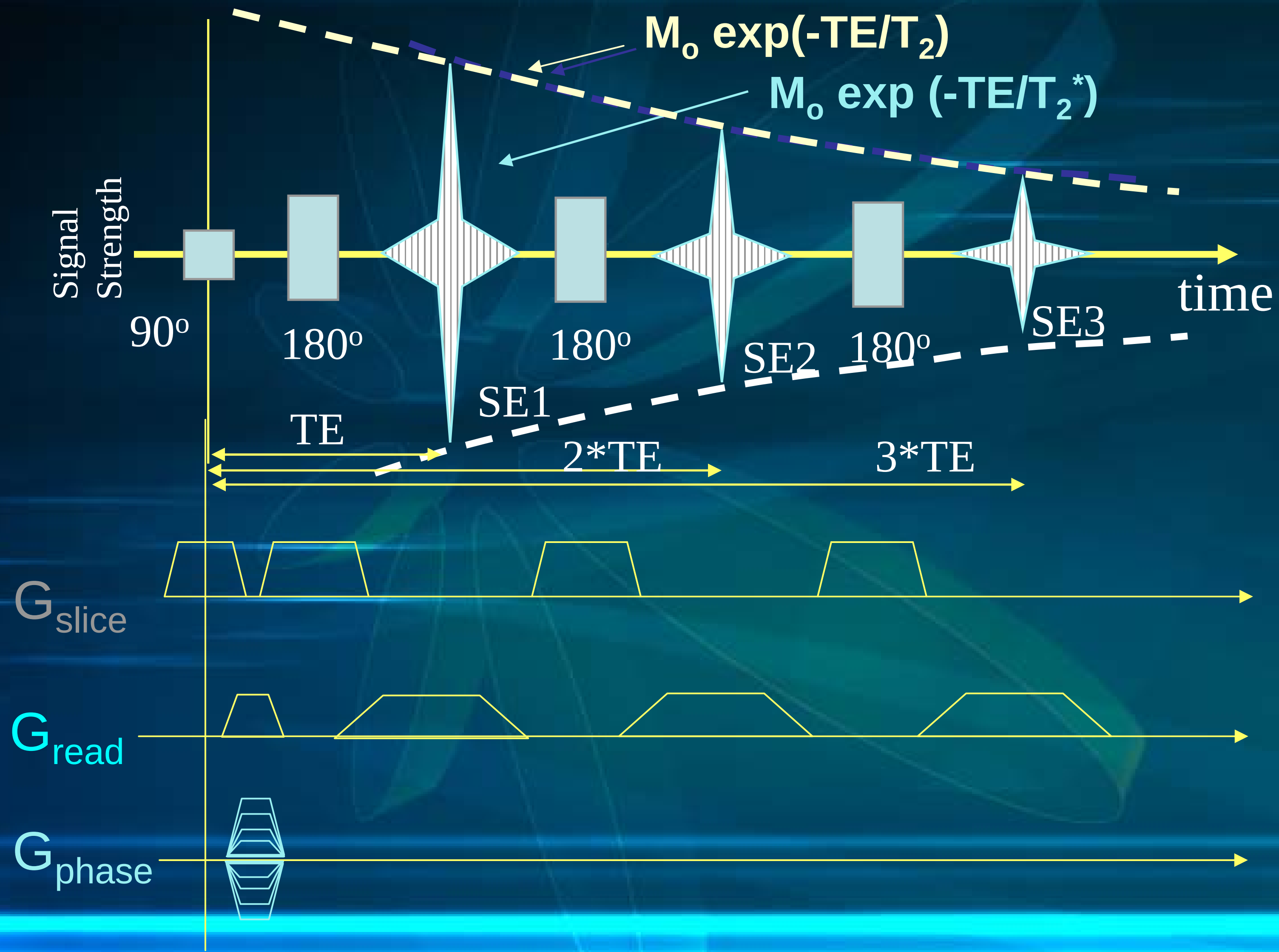
- Never Assume RF Flip Angle is Correct
  - Varies over imaged slice due to slice profile
  - Flip angle must be calibrated across slice
- Be careful in assuming magnetization has reached steady state between acquisitions
- Optimize sequence acquisition parameters to ensure maximal SNR
- Always check that fitted conforms to assumed model

Transverse Magnetization

T2* Decay



Multi-Echo Acquisitions



Calculation of T_2

$$M_{xy} = M_o' e^{-t/T_2}$$

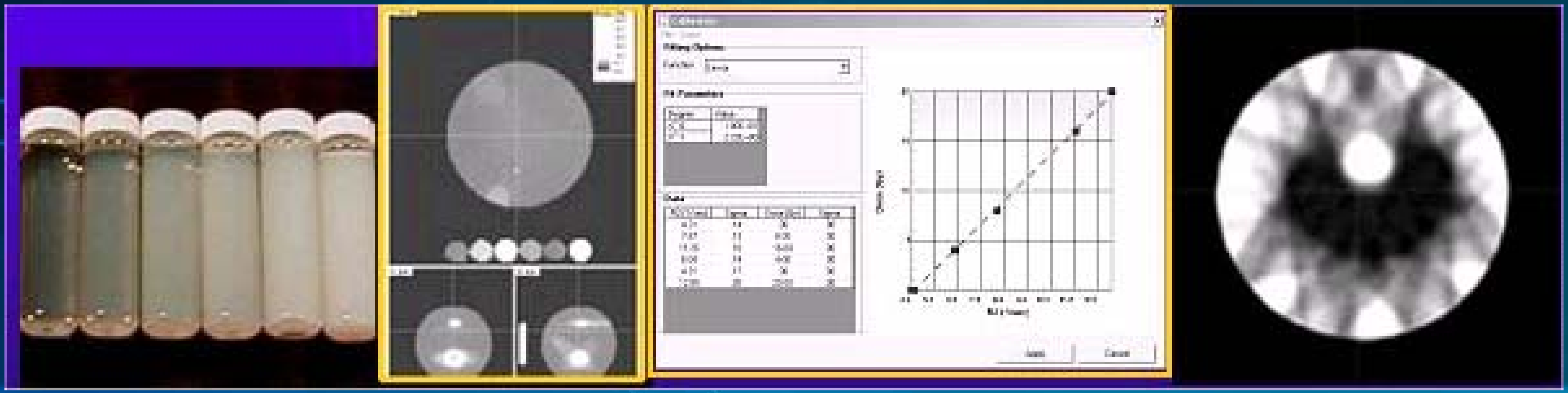
$$\ln M_{xy} = -1/T_2\, t + \ln M_o'$$

$$\ln(M_{xy} / M_o') = slope = -1/T_2$$

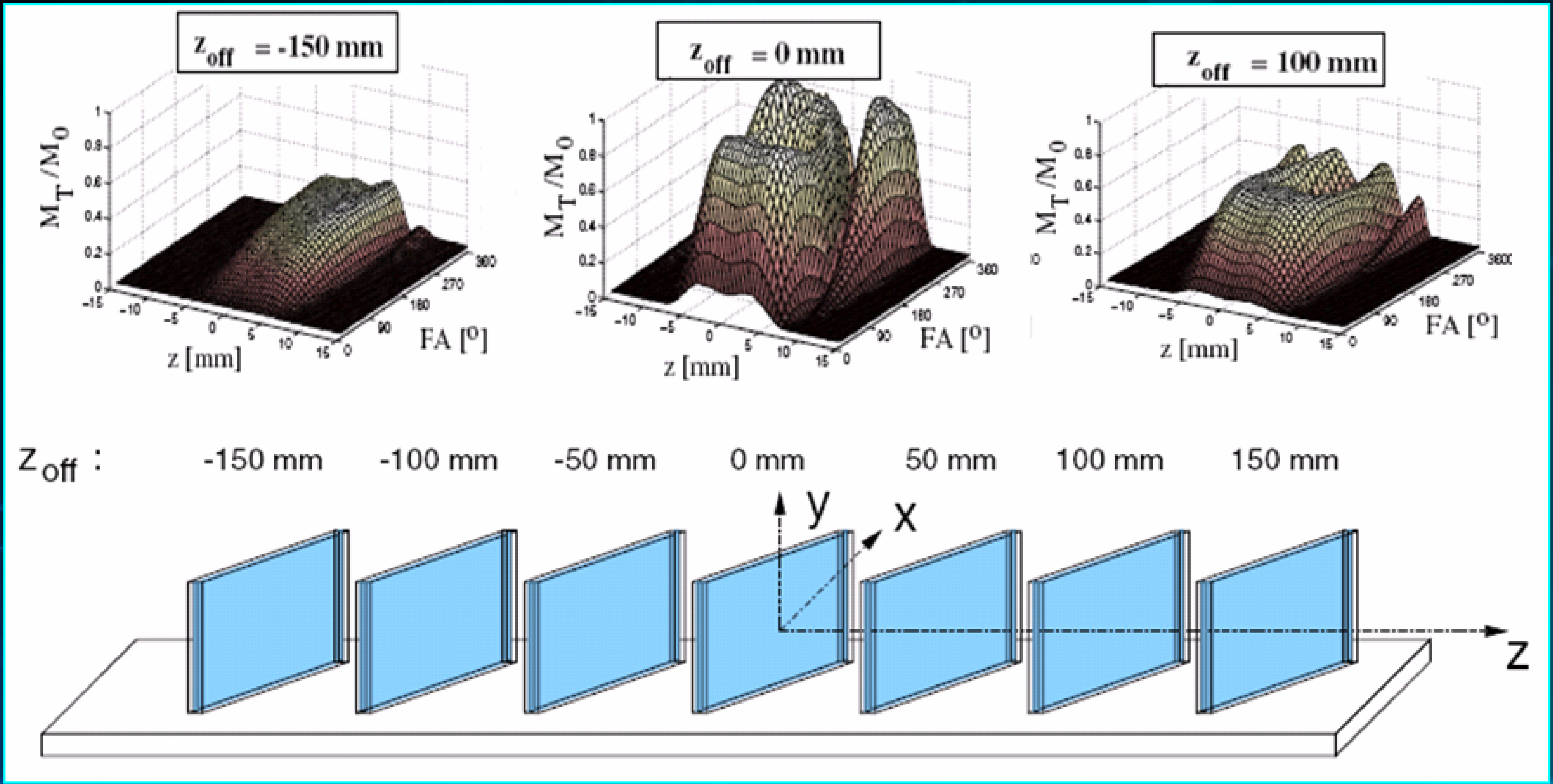
$$T_2 = -1/slope$$

Gel Dosimeters

- Used for 3D Radiation Dosimetry QC
- Relies on direct relationship between relaxation rate, R_2 (R_2=1/T_2) of gel following exposure and dose

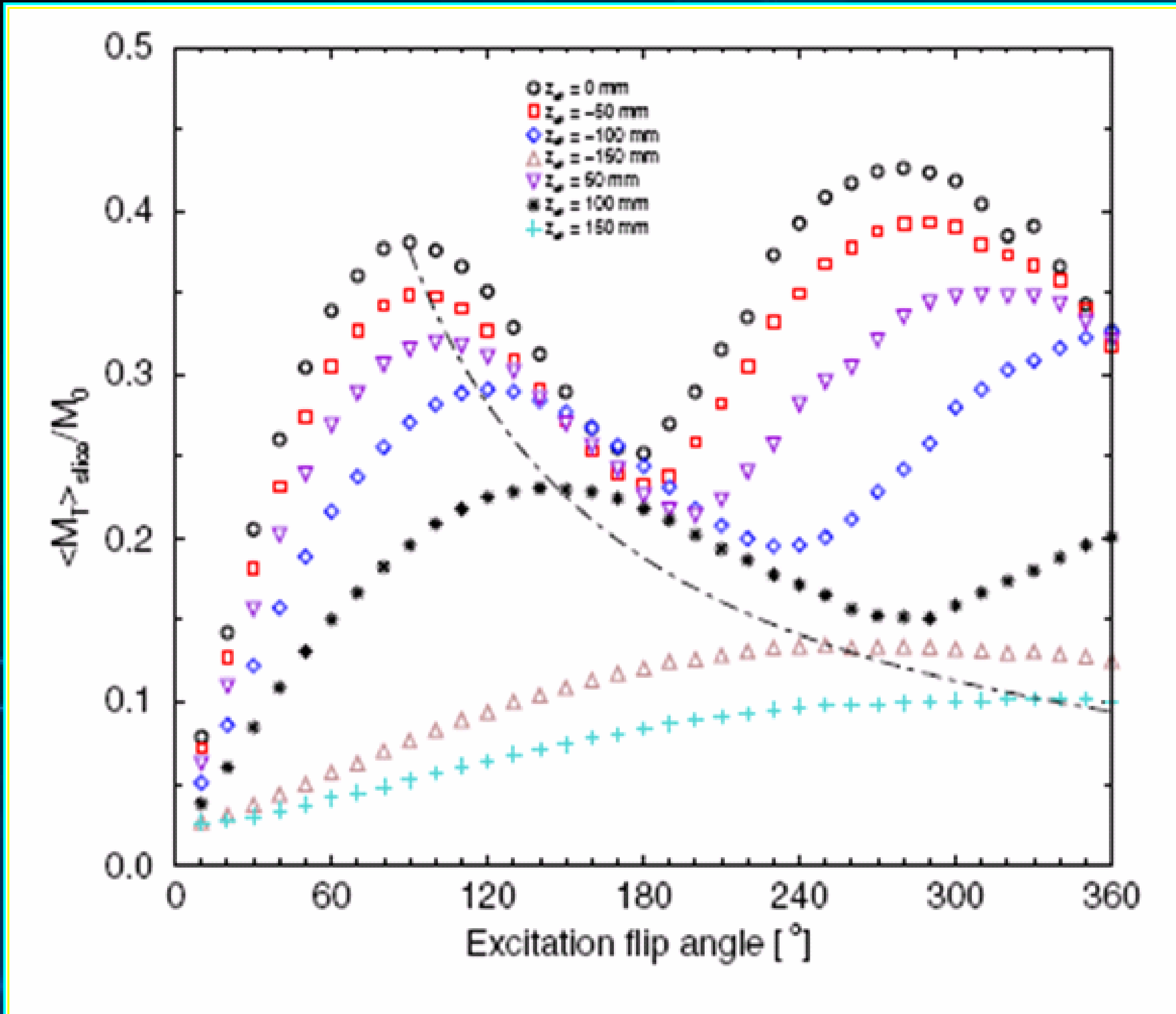


B1 Changes with Slice Position



De Deene Y et a. Phys Med Biol 2000; 45:1825-1839

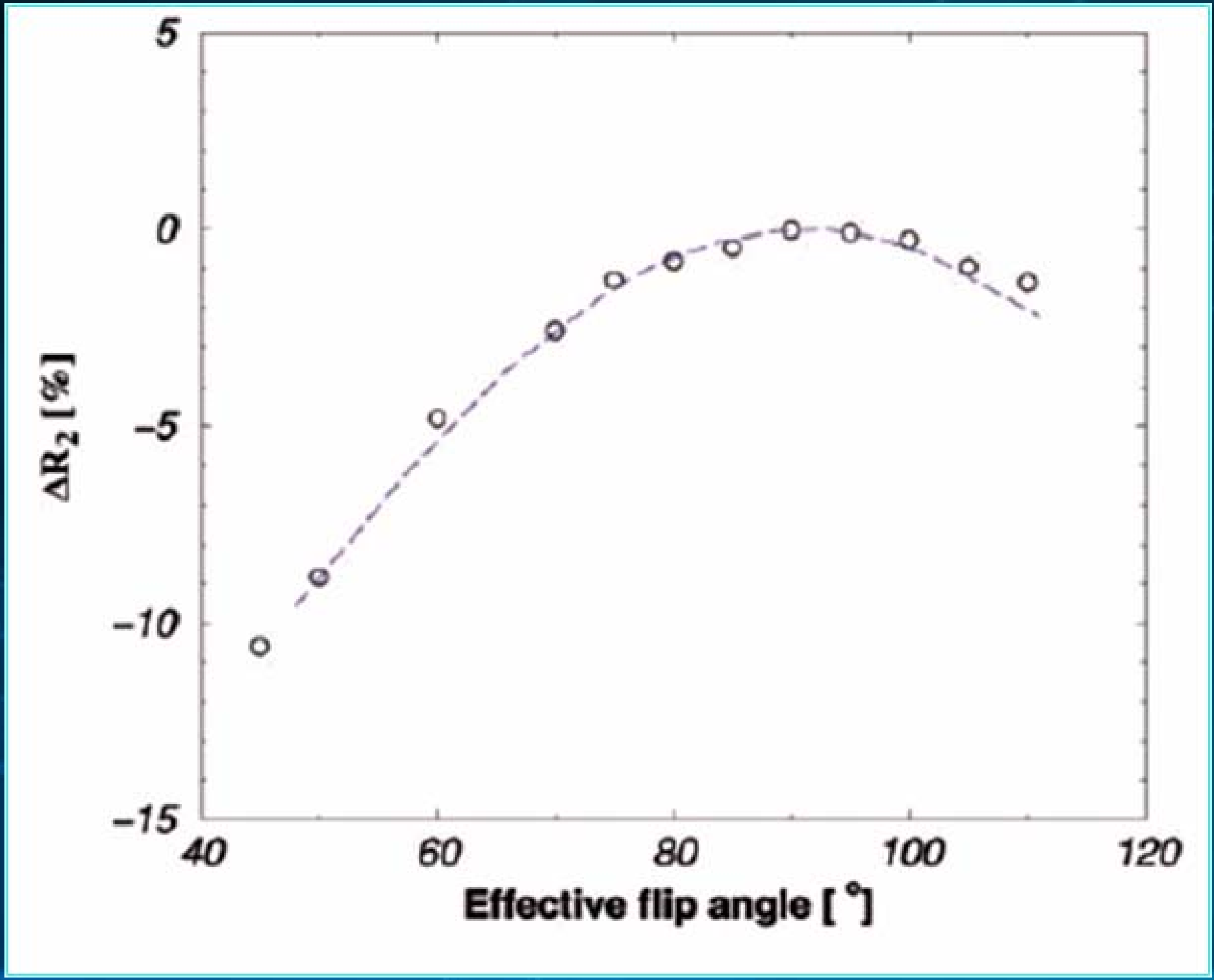
Effective Flip Angles



Average transverse magnetization within a slice as a fraction of M_o for various slice positions for flip angles ranging from 0º to 360º

De Deene Y et a. Phys Med Biol 2000; 45:1825-1839

R_2 Calibration



For Accurate & Precise T2

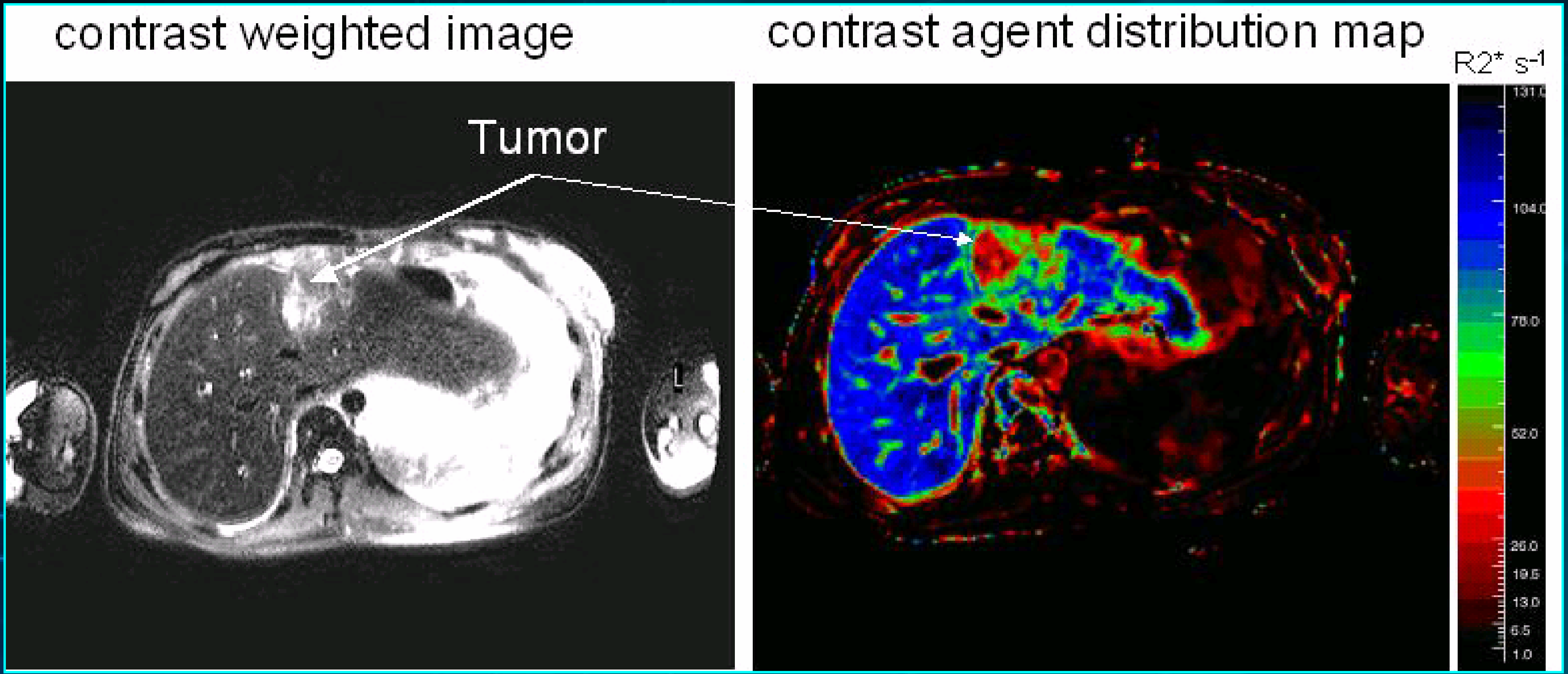
- Never Assume RF Flip Angle is Correct
  - Varies over imaged slice due to slice profile
  - 180^o flip angle must be calibrated across slice
- Use multi-echo (vs. dual echo) approach and big TX coils whenever possible
- Analyze and understand eddy current effects on T2 measurement
- In tissues, beware of multi-exponential decay

T2* Parametric Imaging

$$M_{xy} = M_o' e^{-t/T_2^*}$$

- Similar to T2 measurements but use gradient echo imaging with varying TE

Contrast Agent Maps

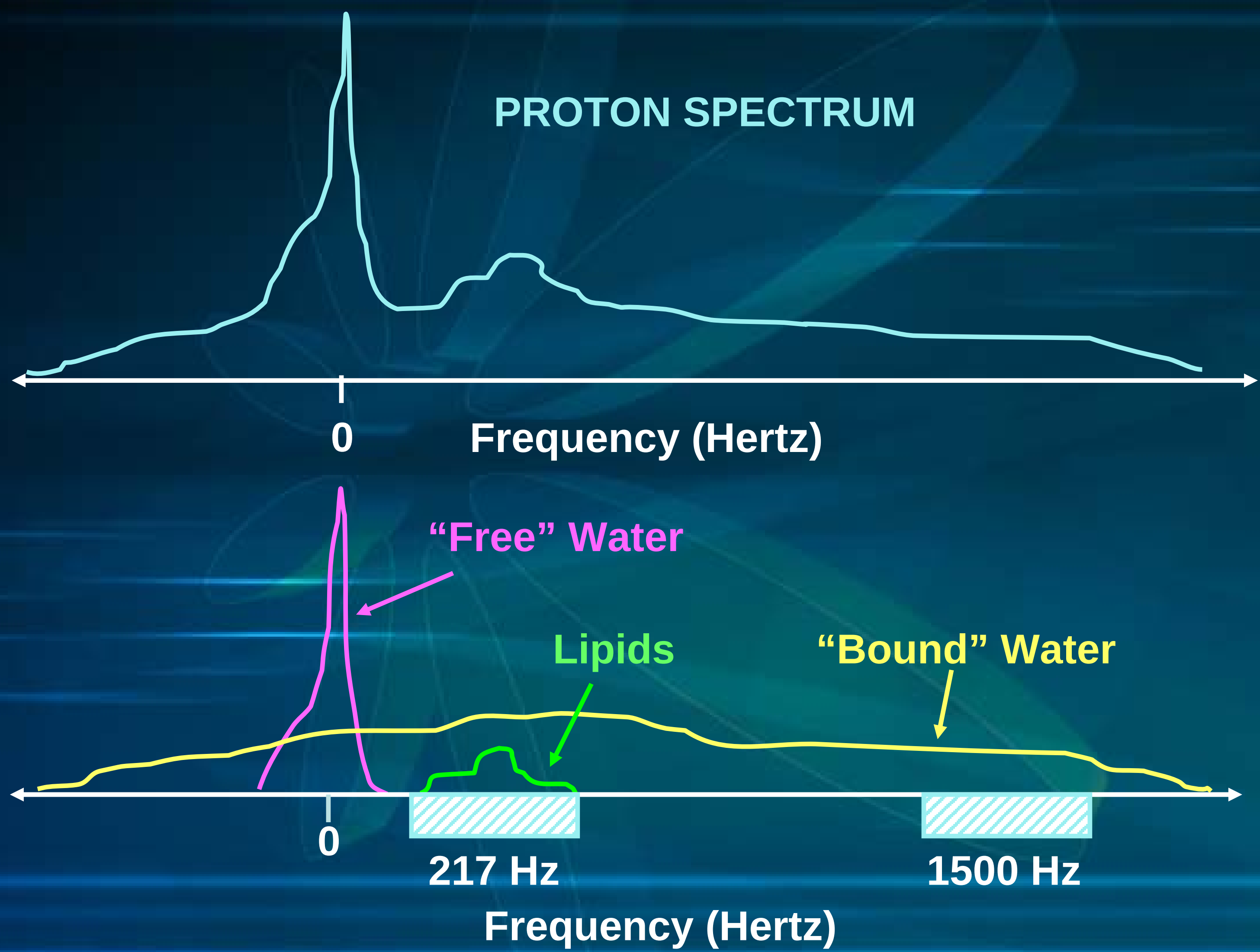


T1-weighted image

Parametric map of R_2^*

http://www.research.philips.com/

Magnetization Transfer



Magnetization Transfer Ratio

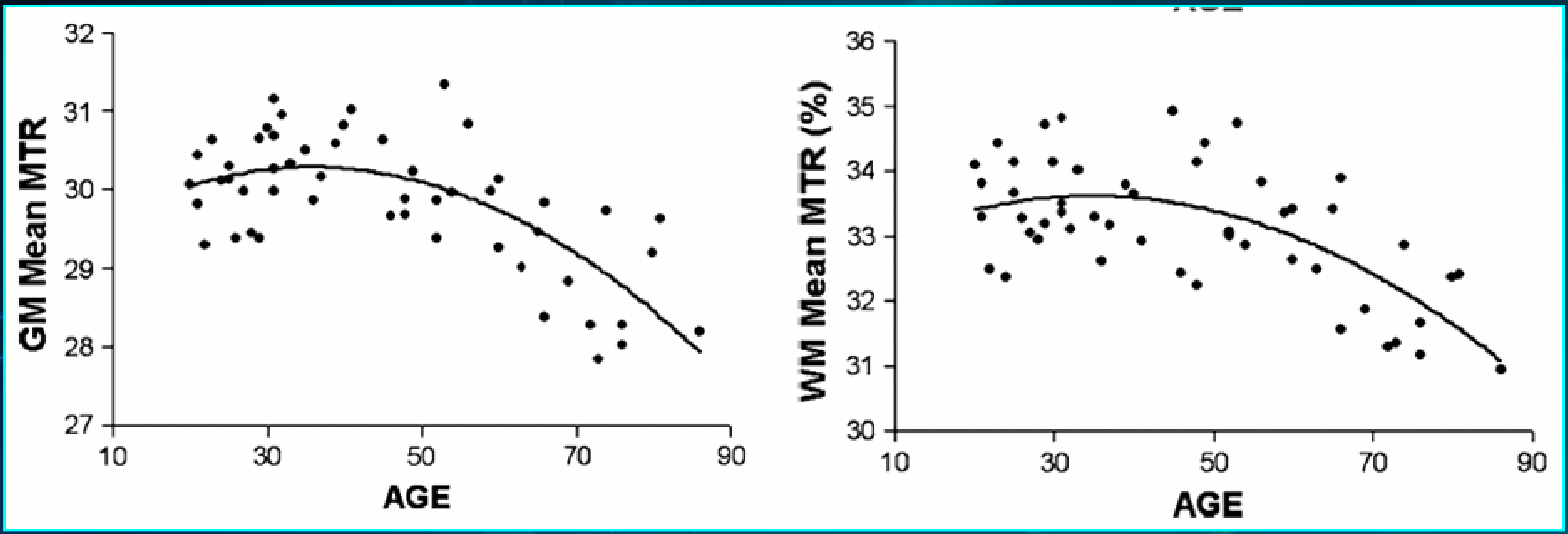
$$MTR = \frac{M_o - M_s}{M_o}$$

Magnetization Transfer Ratio (MTR)

- the difference of the saturated versus non-saturate images relative to the signal in the normal (non-saturated images)

MTR and Aging

Gray matter and white matter MTR images reveal a quadratic change with age that is primarily attributed to normal demylenation



Inglese & Ge, Top Magn Reson Imag 2004; 15:355-363

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Physiological Measurements

- Flow – bulk motion of blood and other fluids within body
- Perfusion – amount of blood traveling through capillaries in ml/s/gm of tissue
- Diffusion – random motion of spins in a homogeneous solution
  - Apparent Diffusion Coefficient – measured diffusion rate of water through tissue

Attenuation Due to Diffusion

$$A(TE) = A(0)\exp[-\gamma^2 G^2 D_{app}\delta^2\alpha^2(\Delta - \frac{\delta}{4})]$$

Where: $\alpha=\pi/2$;

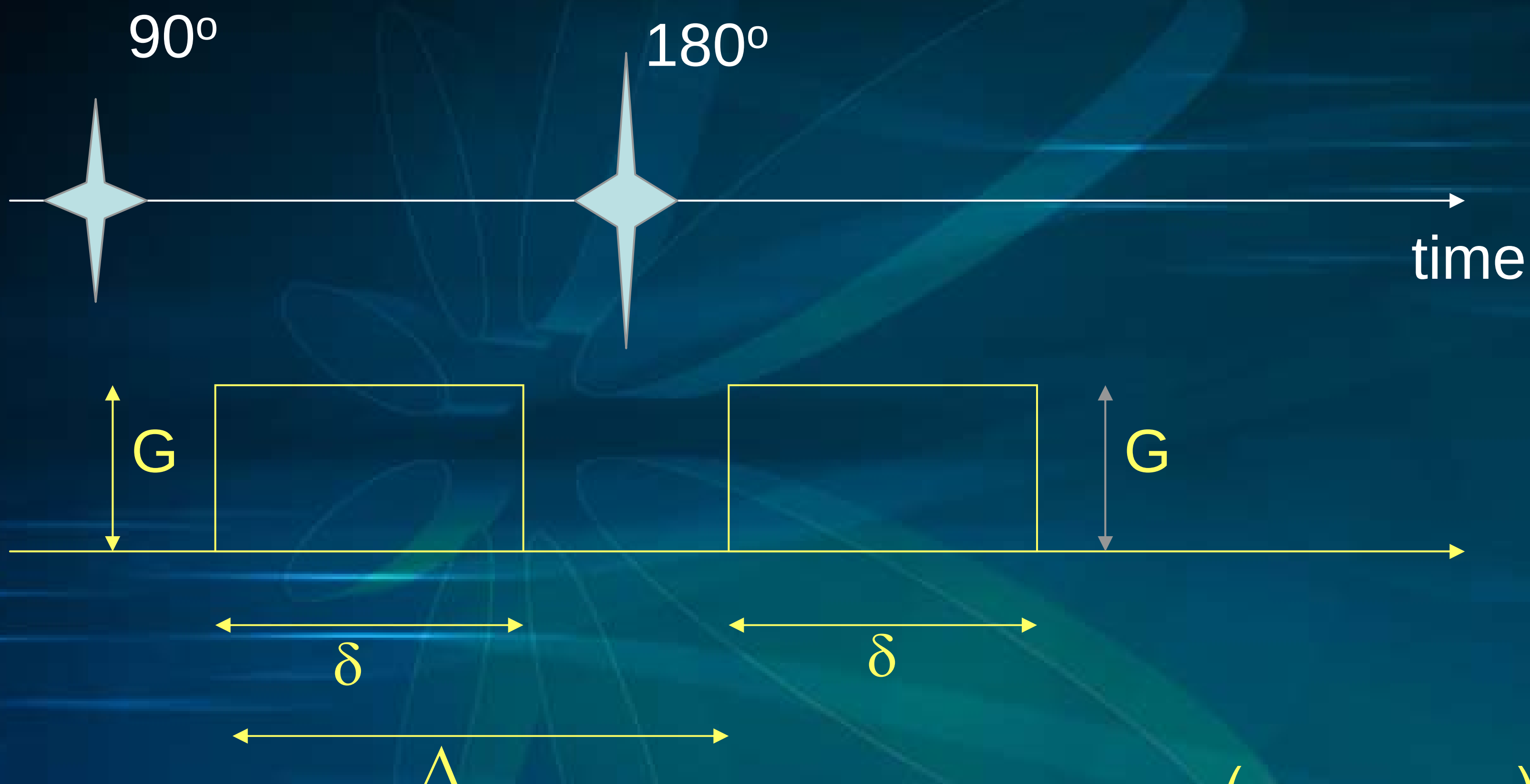
G is amplitude of diffusion sensitive gradient pulse;

δ is duration of diffusion sensitive gradient;

Δ is time between diffusion sensitive gradient pulses;

D_{app} is the apparent diffusion coefficient

DWI Basic Pulse Sequence



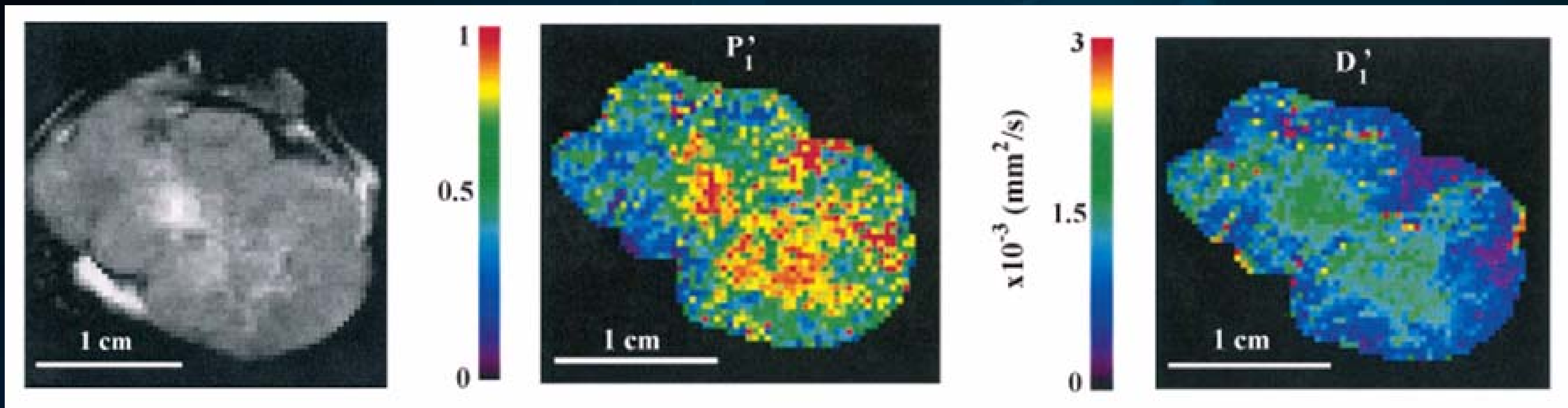
$$b = \gamma^2 G^2 \delta^2 \left(\Delta - \delta/3\right)$$

Stejskal EO & Tanner JE, J Chem Phys 1965. 42: 288-292

The b-value

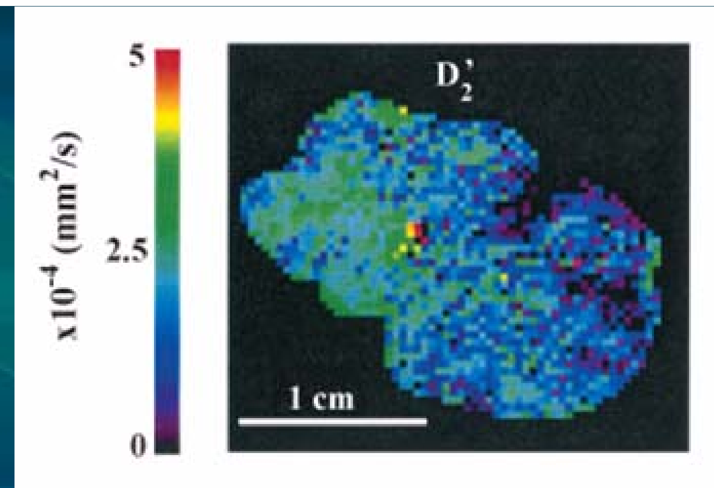
- Controls amount of diffusion weighting in image
- The greater the b-value the greater the area under the diffusion-weighted gradient pulses
  - longer TE
  - stronger and faster ramping the gradients

DW MRI on Breast Tumor



Parametric maps calculated from biomodal exponential decay model:

$$\frac{I}{I_o} = P_1^{'} \exp\left(-bD_1^{'}\right) + P_2^{'} \exp\left(-bD_2^{'}\right)$$



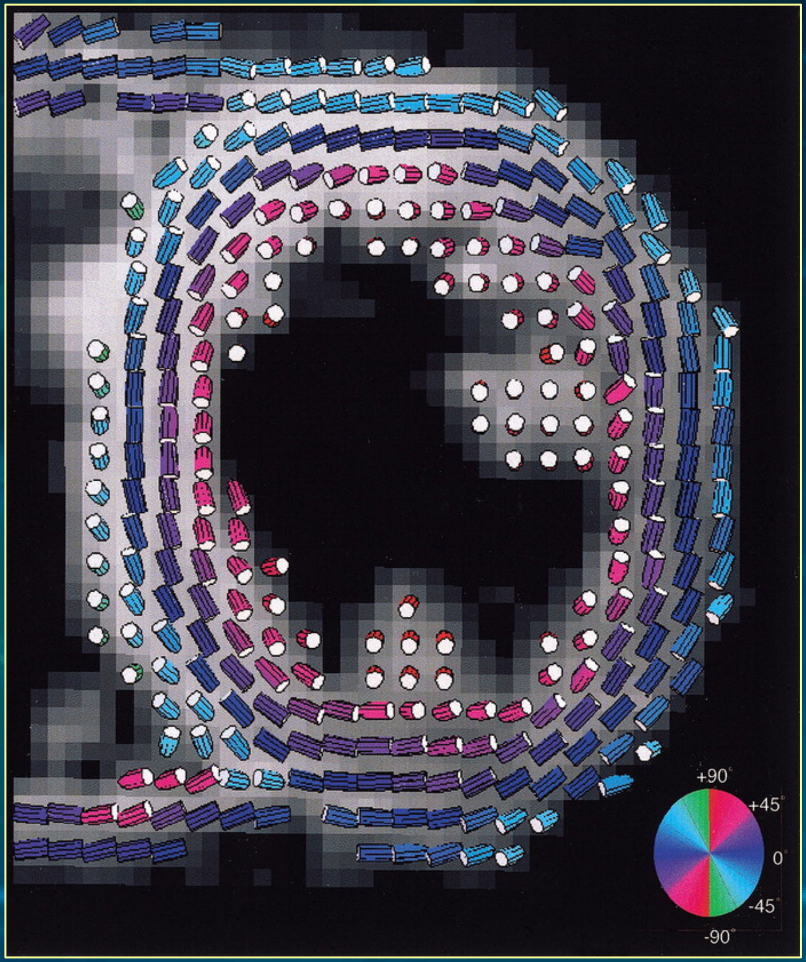
Paran Y et al. NMR Biomed 2004; 17:170-180

Anisotropic Diffusion

Restricted diffusion along neural fibers

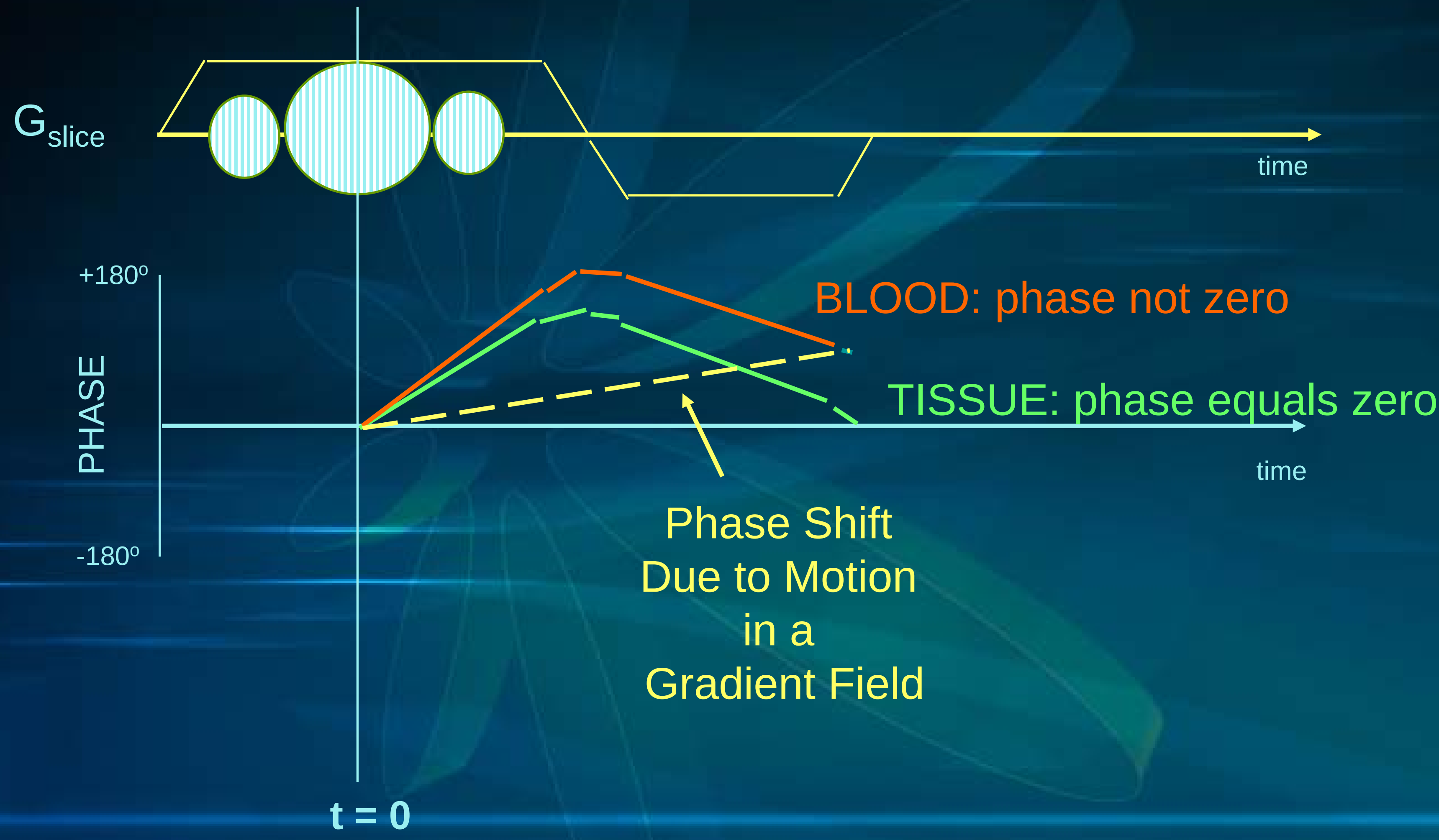
Diffusion Tensor Imaging

- In anisotropic tissues (neural fibers, muscle fibers) scalar ADC depends on direction of diffusion sensitizing gradient
- Diffusive transport of water can be characterized by an *effective* diffusion tensor
- Direction of diffusion can be used to create a map showing orientation of myocardial fibers.

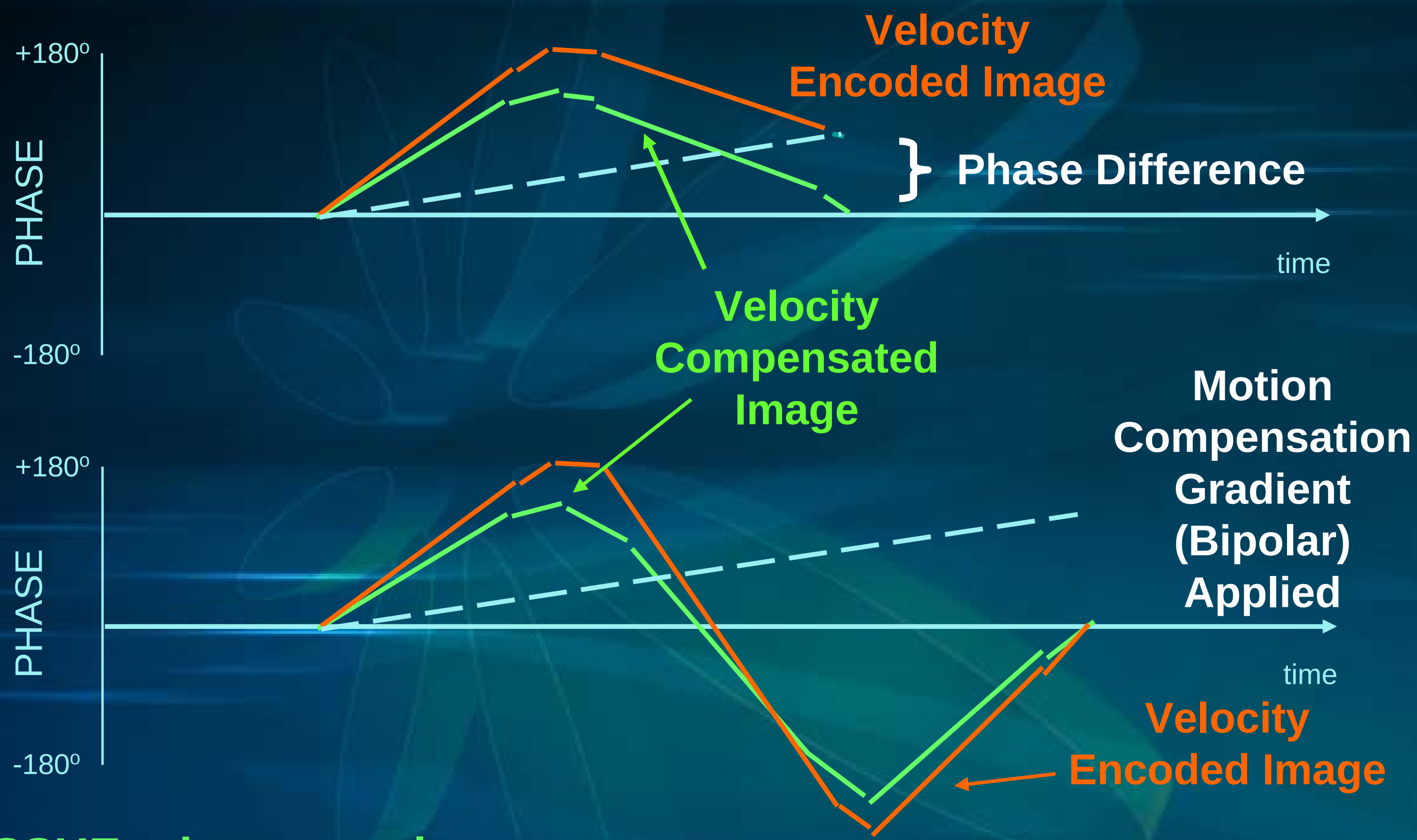


Tseng et al. , Radiology 2000; 216: 128-139.

Dephasing Due to Motion



Phase Contrast Imaging



TISSUE: phase equals zero in BOTH images

BLOOD: phase is DIFFERENT in each image

Phase Contrast Images

Two Signals:

$S_1 = S_s + S_m$;$S_2 = S_s + S_m\ e(i\varphi_m)$

where $\varphi_m = \gamma\ \Delta M_1\ \upsilon$ (υ = velocity)

Complex difference -

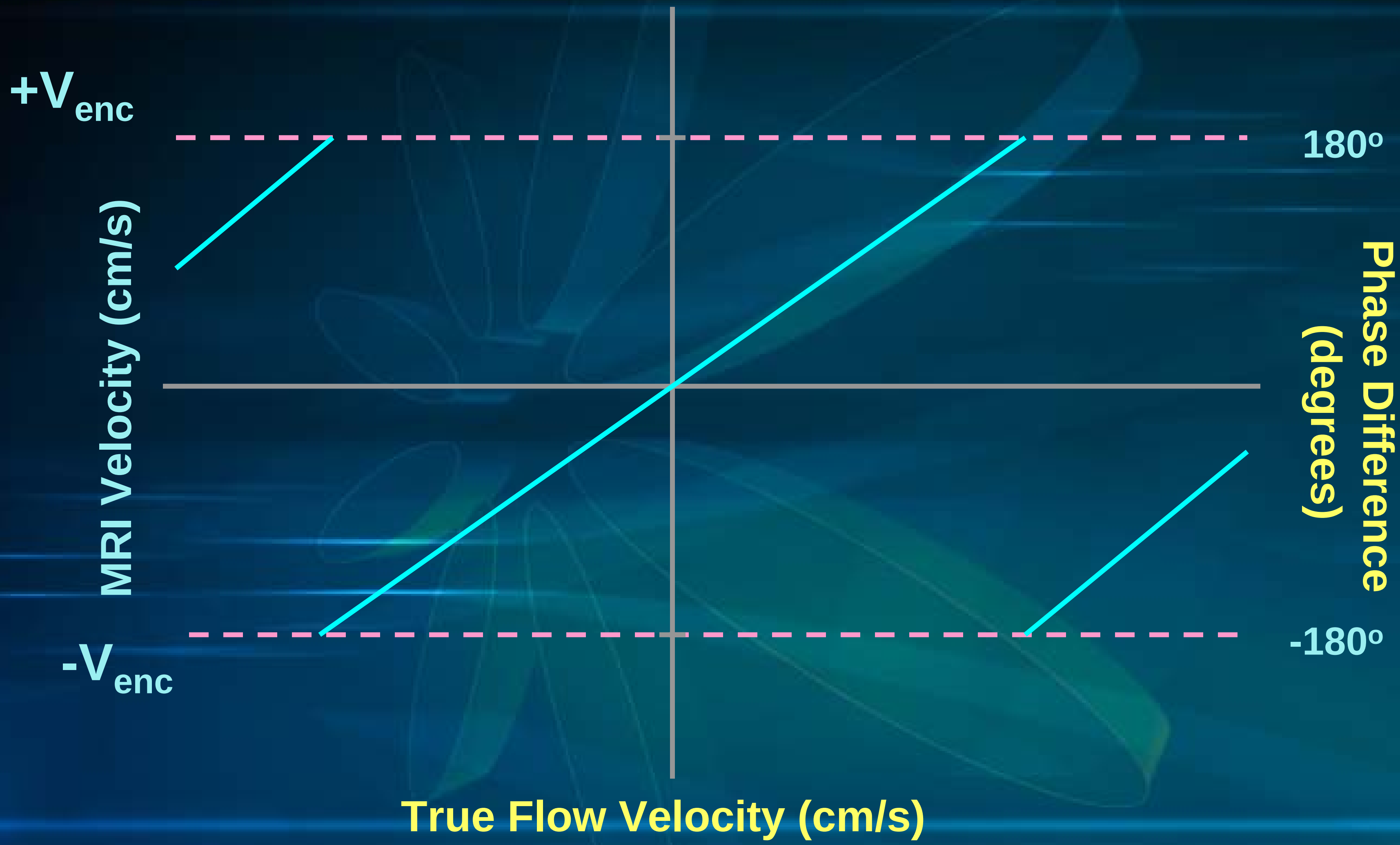
$\Delta S = S_2 - S_1 = S_m[e(i\varphi_m)-1] = 2iS_m \sin(\varphi_m/2)$

Phase difference -

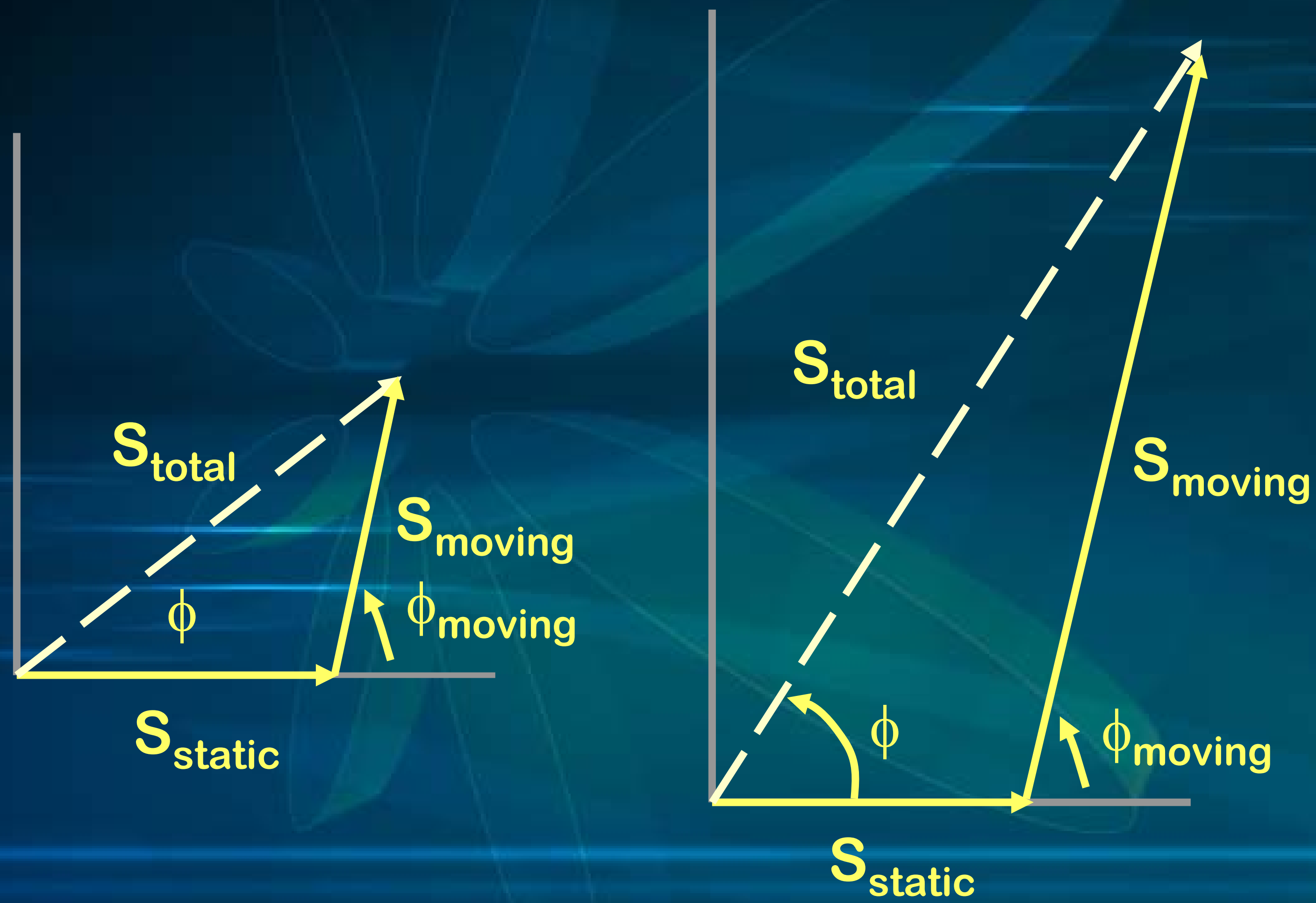
$\Delta\varphi = \arg S_2 - \arg S_1$, then

$\upsilon = \Delta\varphi / (\gamma\ \Delta M_1)$

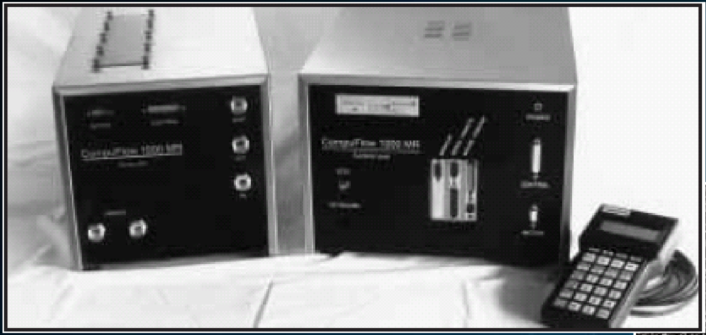
Velocity Encoding (V_{enc})



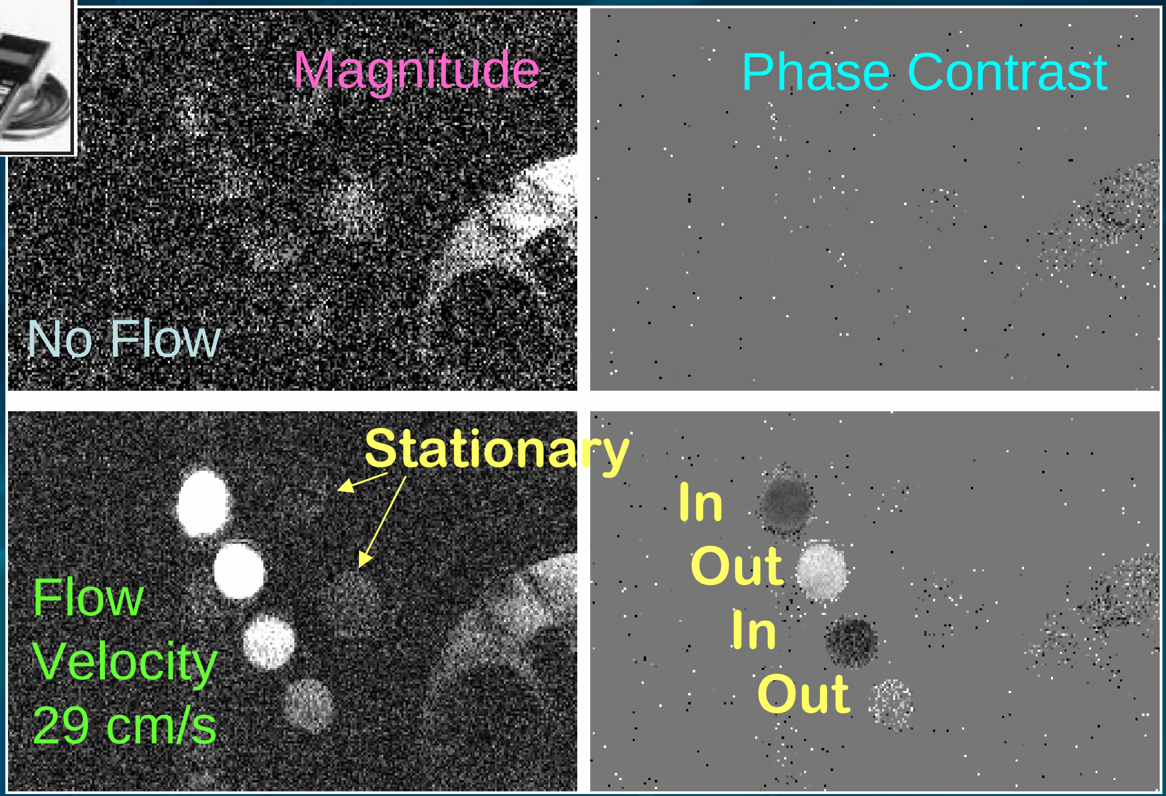
SNR in Phase Contrast



Flow Phantom Calibration

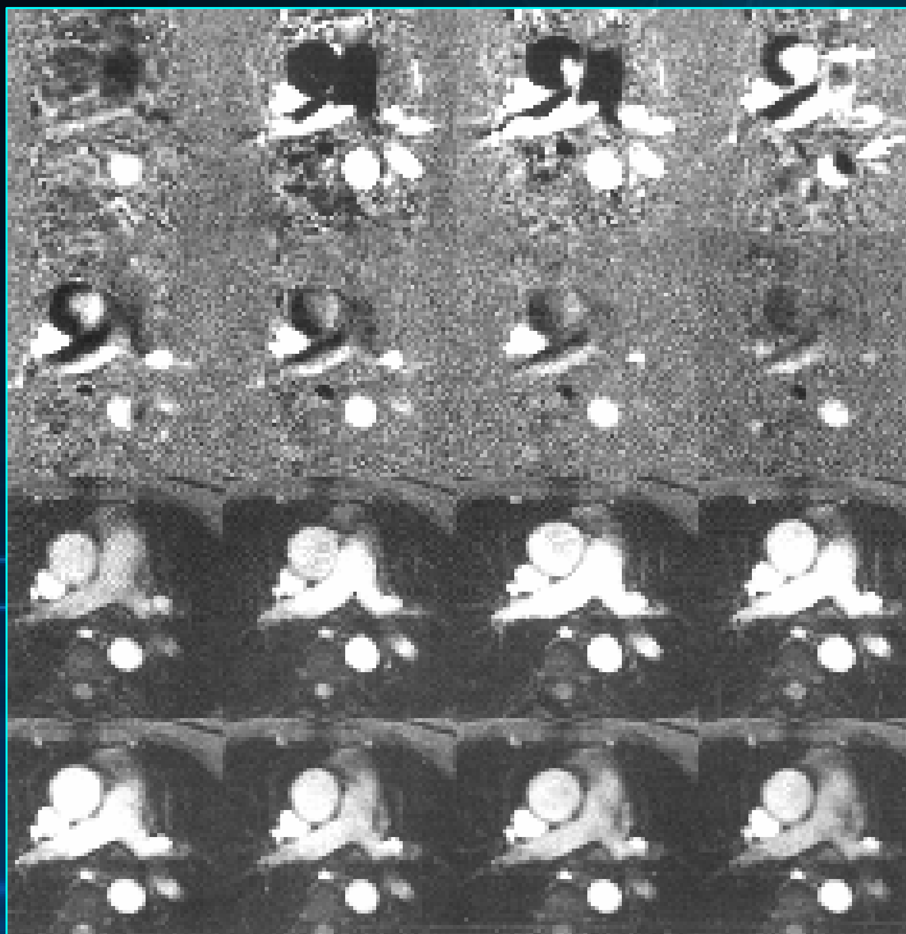


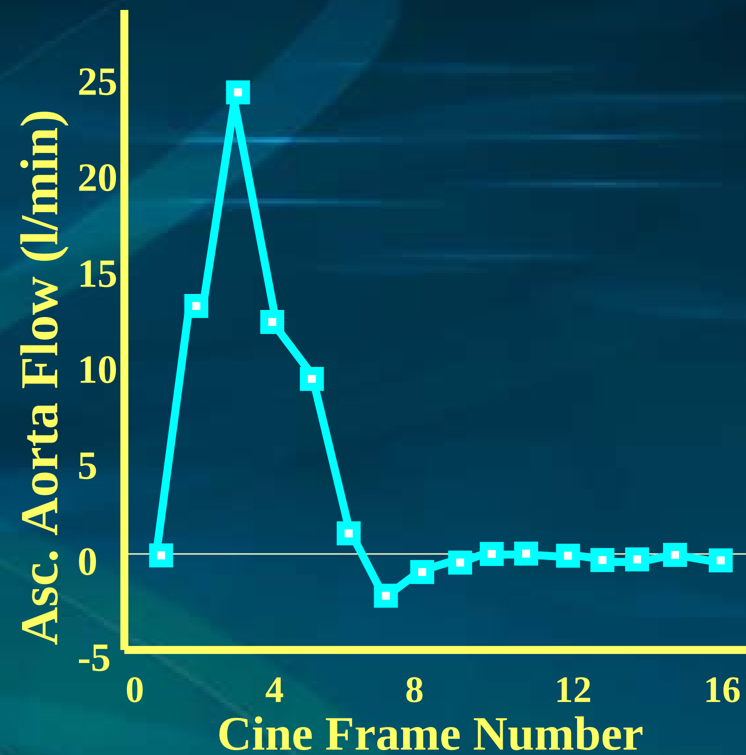
Can use commercially available flow pumps to accurately simulate blood flow in various vessel in the body.



http://www.simutec.com/

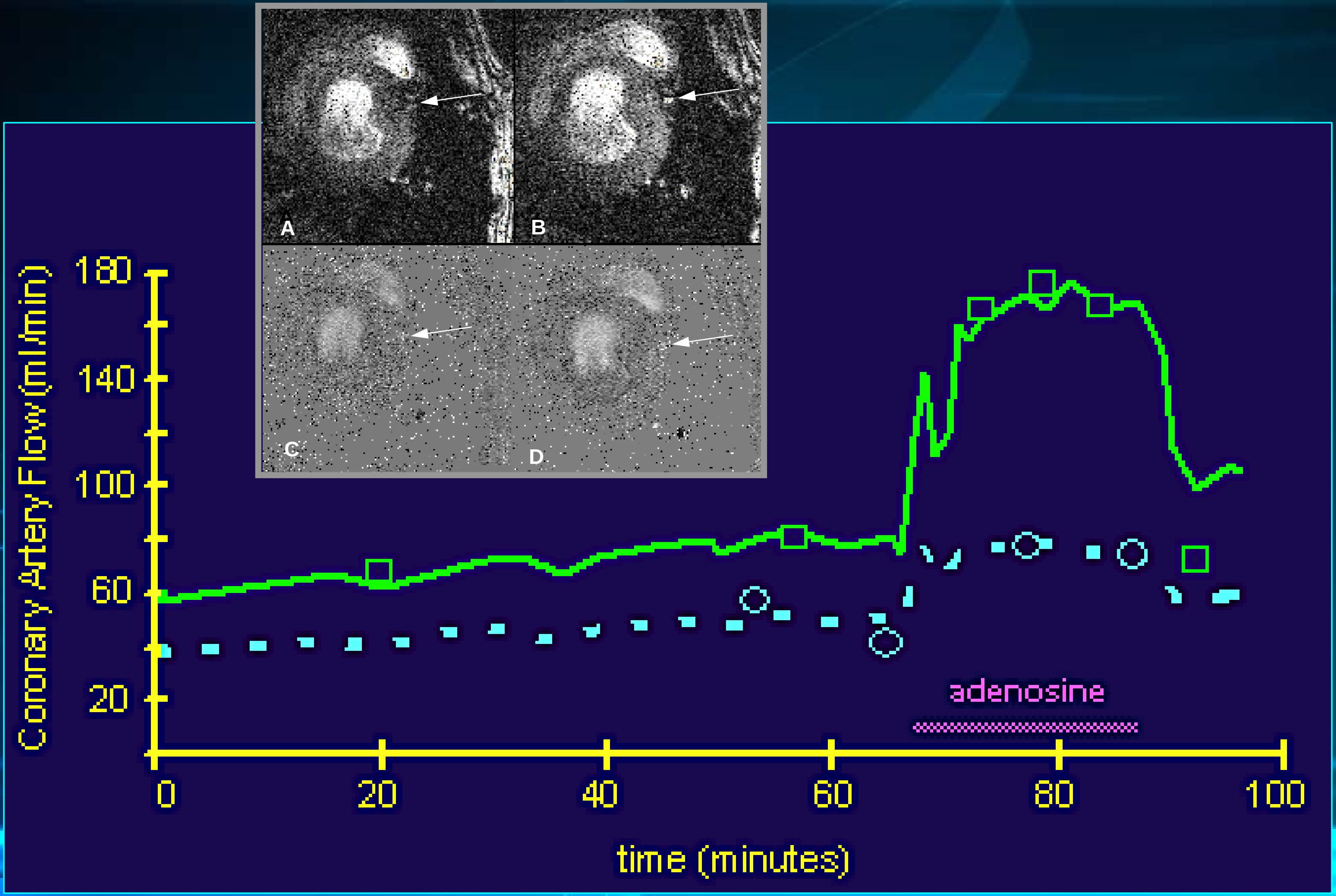
Cardiac Output



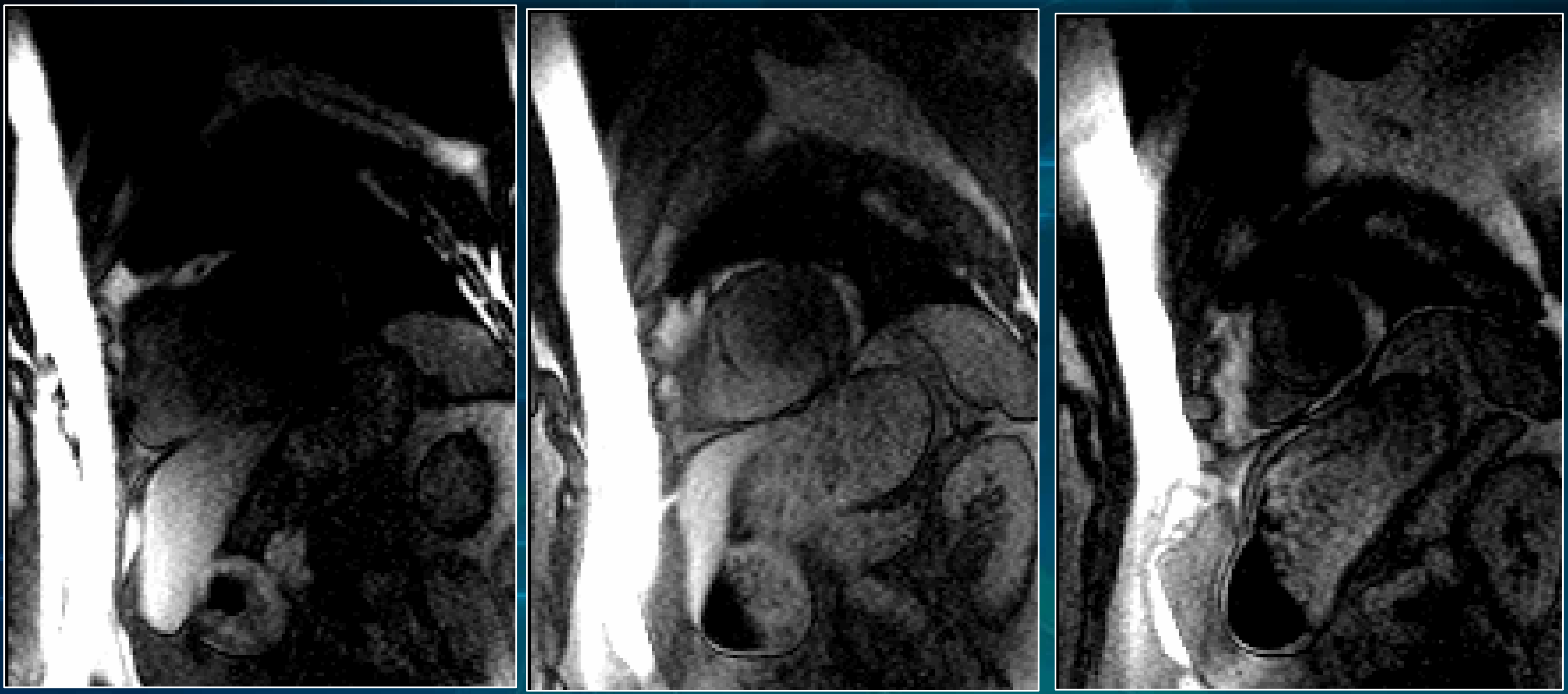


Cardiac Output = 4.01 L/min
Stroke Volume = 68 ml

Coronary Artery Flow



Myocardial Perfusion



Short-axis views of patient's heart showing Gd-DPTA Uptake into RV, LV and then myocardium

Nonuniformity Correction

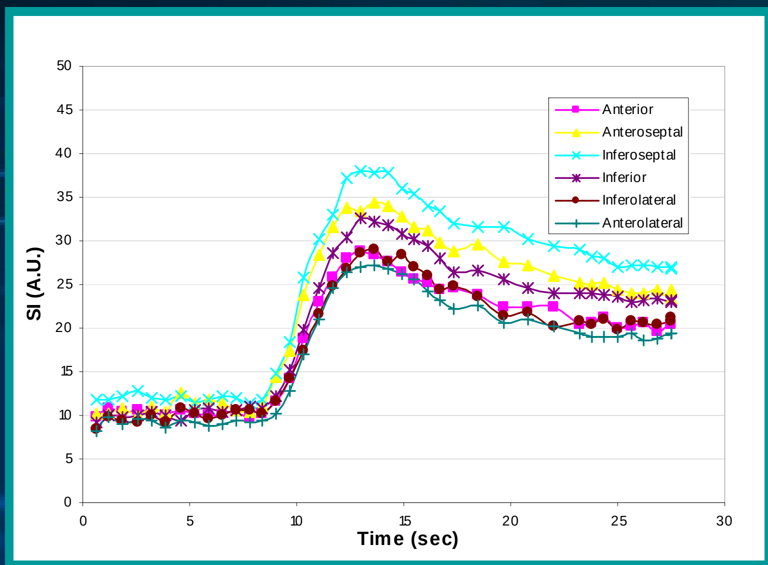
Real-time TrueFISP (GRAPPA=2) cine study was performed between two perfusion scans on the heart of a normal volunteer

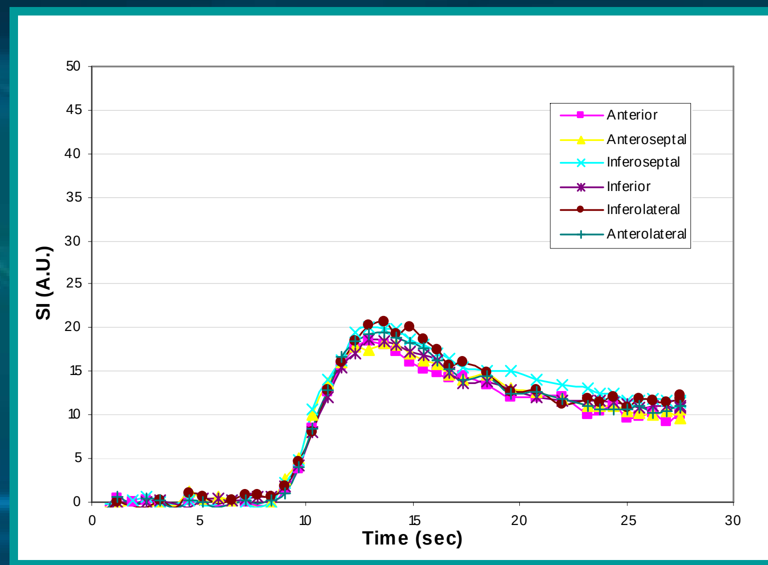
$$SI_{normalization} = SI_{perf}/SI_{cine}$$

$$SI_{correction} = SI_{normalization} - SI_{base}$$

Correct for overall effects

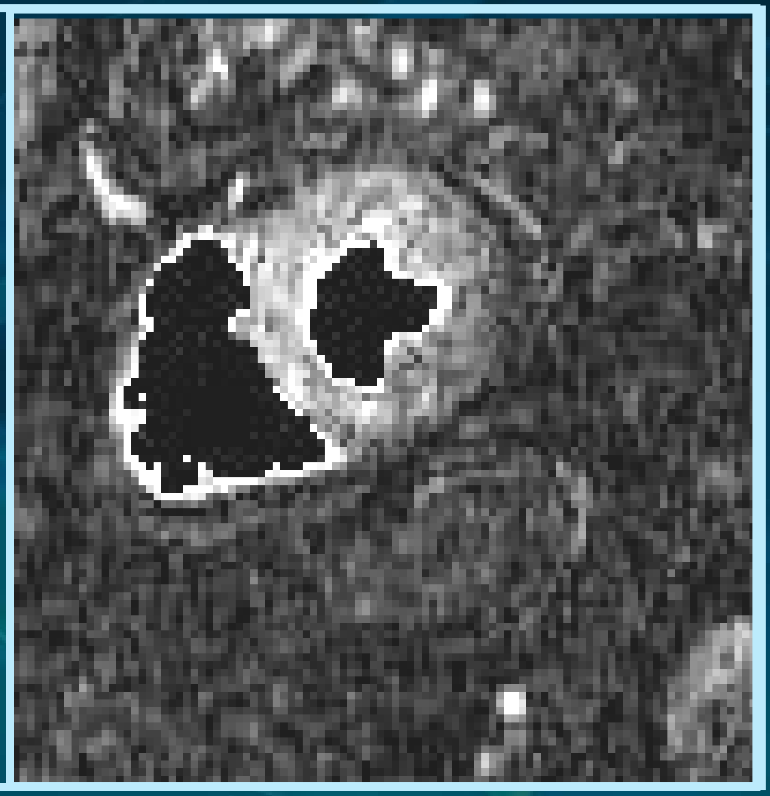
Correct for baseline signal





Max Upslope Parametric Map

MRI perfusion image



Maximum upslope parametric map

SUMMARY

Before undertaking qMRI:

- Check gradient calibrations
- Understand gradient non-linearities
- Evaluate eddy currents
- Measure RF pulse changes in space
- Determine RF receive nonuniformities

Quantitative MRI Methods

- Measuring things with MRI:
  - Diffusion Coefficients
  - Frequency of a signal
  - Relaxation rates (T1, T2 , MTC)
  - Velocity of motion
  - Volumes of tissues

Learn More Details ...