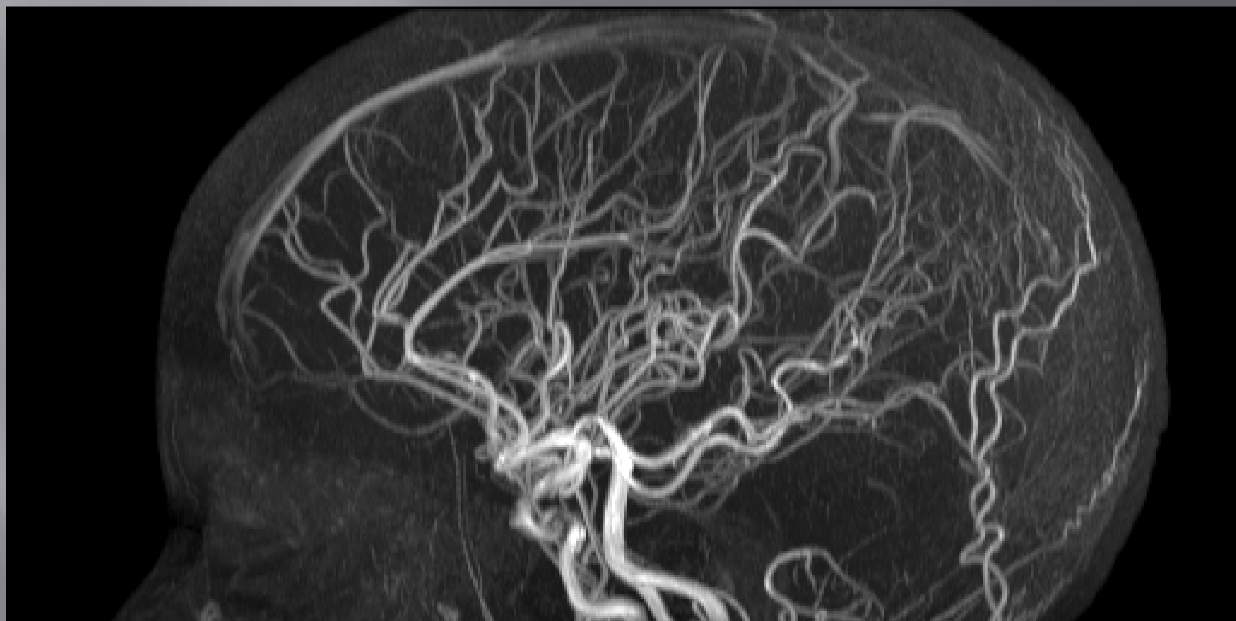
Gradient echo or T2 weighted imaging: An introduction and some clinical applications*



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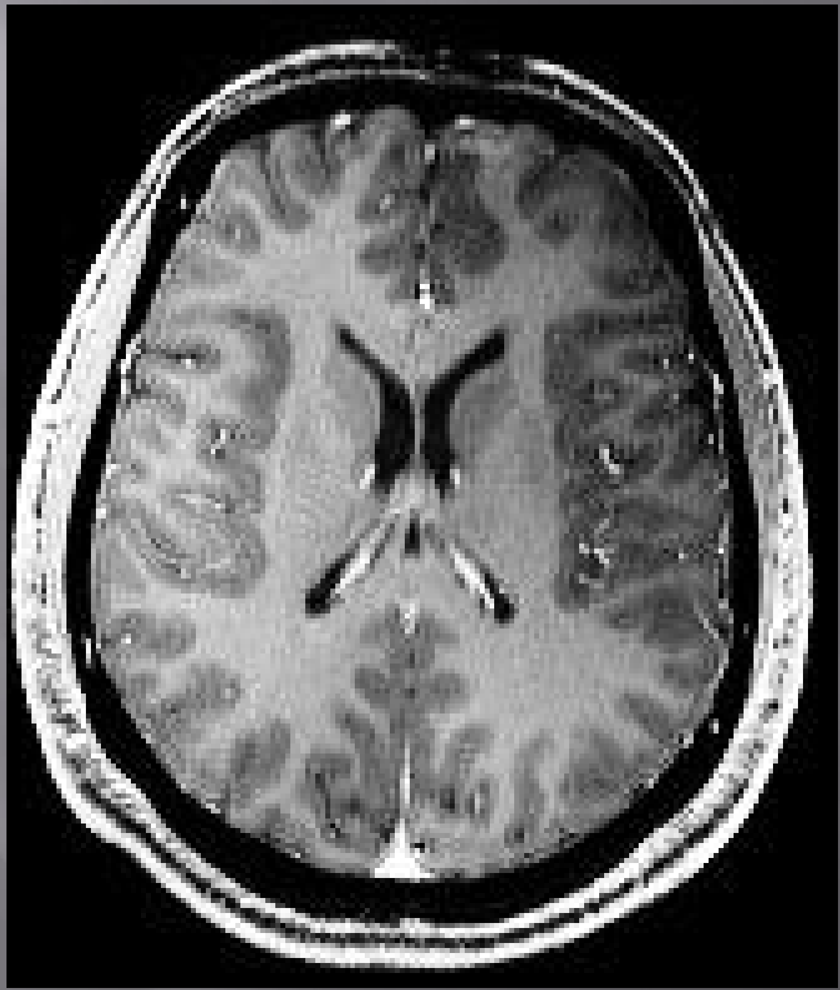
Detroit, Michigan

MAGNETIC RESONANCE PIONEERS

By 1946 Bloch and Purcell and co-workers captured the essence of the behavior of an atom with a non-zero magnetic moment situated in a magnetic field. Their contributions to the field from these first papers was precocious and even 70 years later their work carries modern significance.

Although quantum physics was needed to formulate the experimental and theoretical results, the final impact rests on the very simple relationship given by the Larmor equation relating the frequency of rotation or precession frequency (ω) and the local magnetic field (B) via $\omega = \gamma B$.

Visualizing Quantum Mechanics

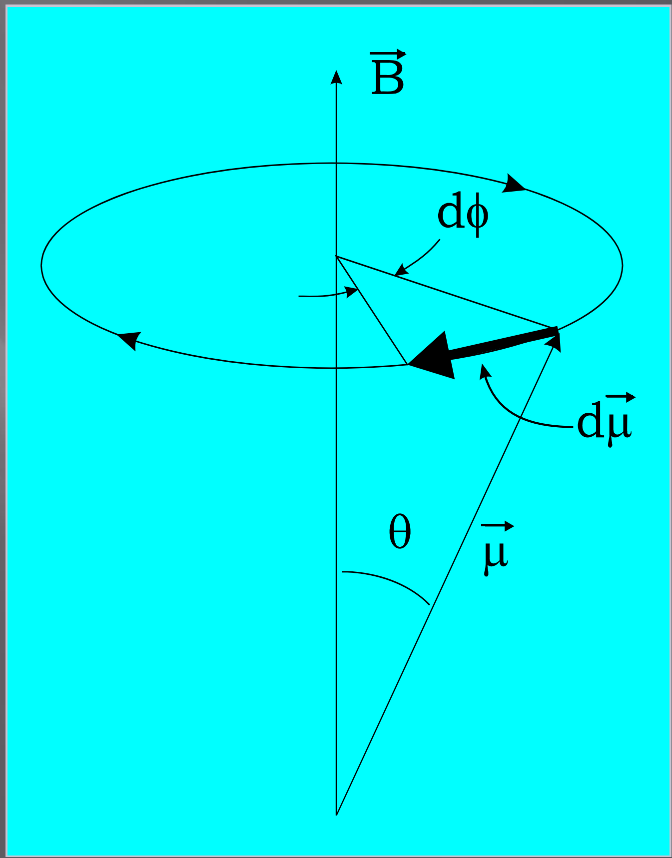


Thin slice T1 weighted post contrast 3D MRI of the brain

Basic MRI concepts

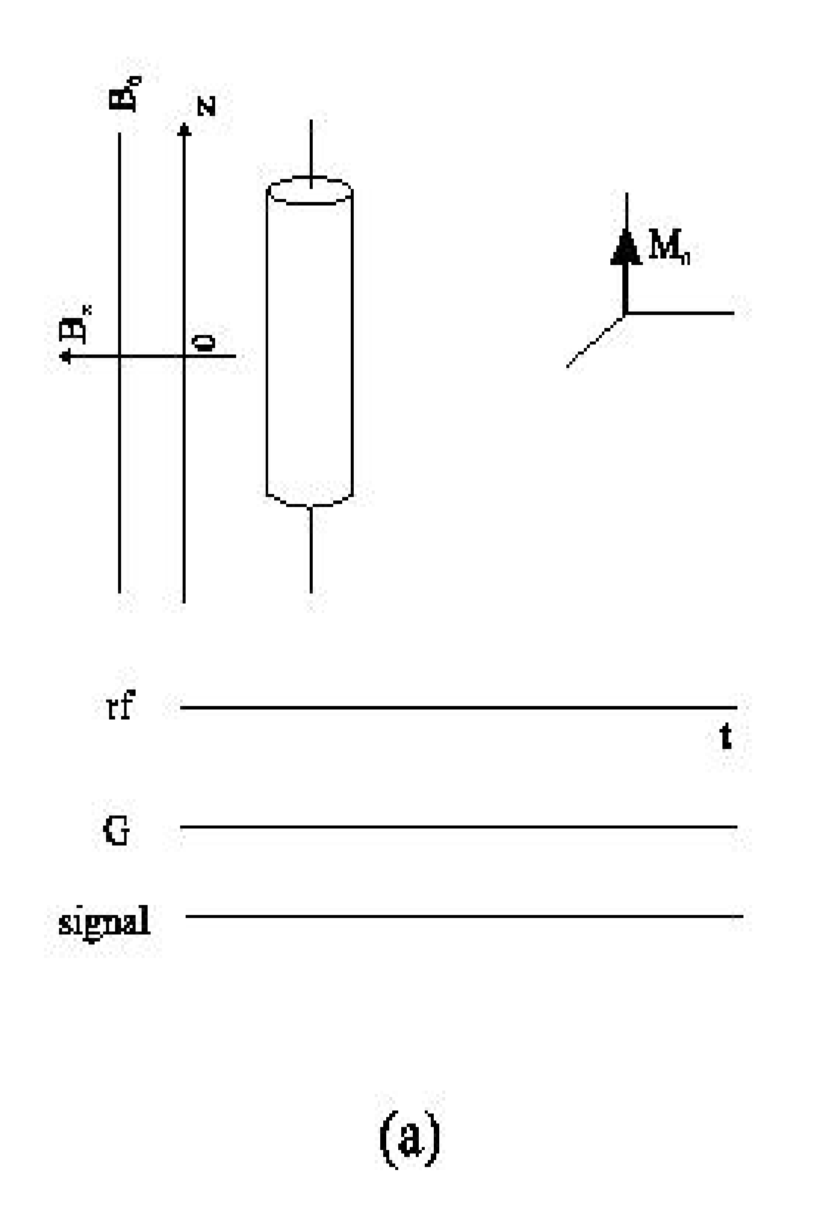
$$\omega \;=\; -\gamma * \mathbf{B}$$

- **B** the applied magnetic field
- γ the gyromagnetic ratio for the proton
- ω the magnetic resonance frequency



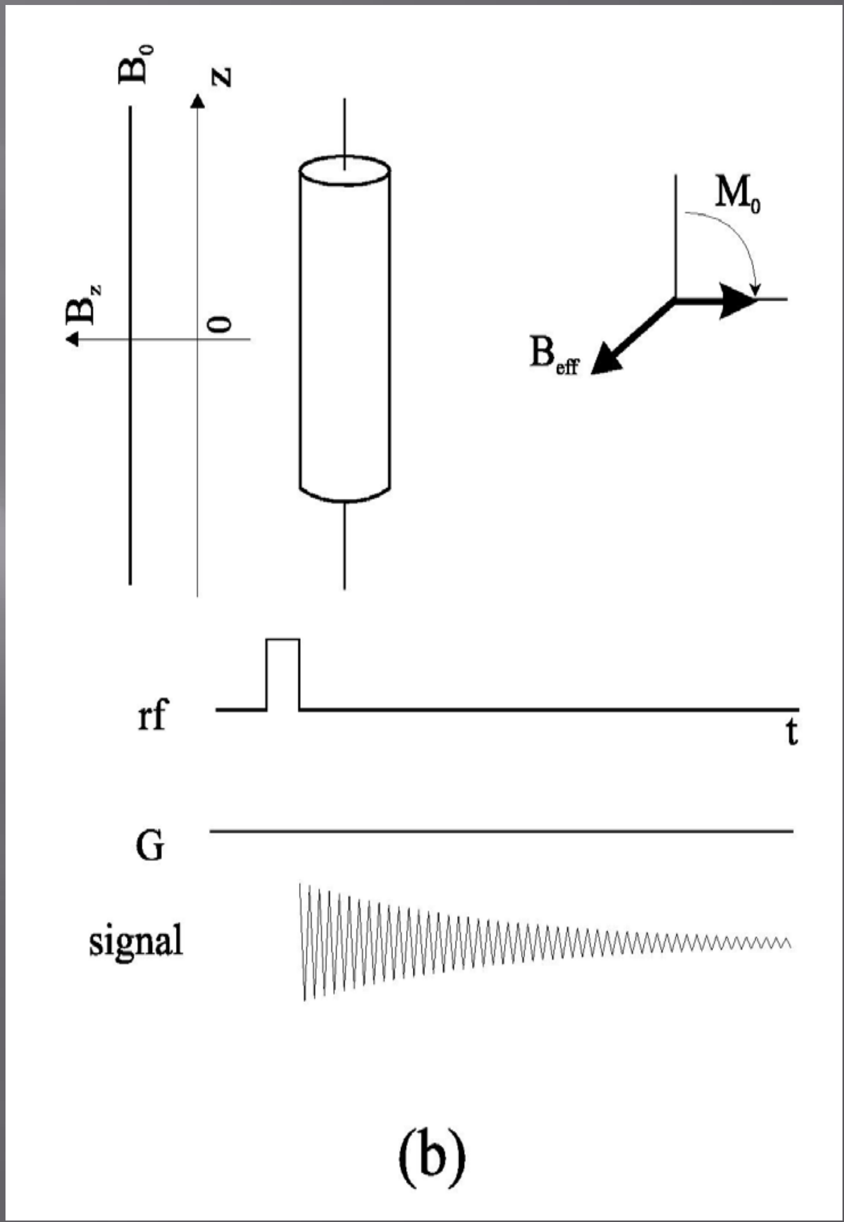
A cylindrical object in a constant magnetic field.

Its bulk magnetization lies parallel to the main field.



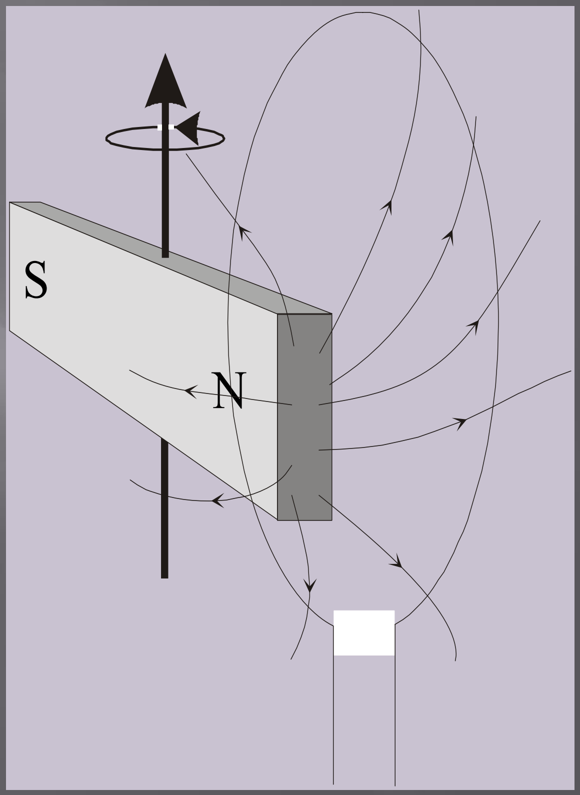
Applying an oscillating rf field along the x-axis rotates M_0 from the z-axis into the transverse plane so that it lies along the y-axis.

At this time, the rf field is turned off and the transverse magnetization precesses about the main field B_0.



MR Signal Readout

The precessing transverse component induces a voltage in a coil which is placed so that it sees a changing magnetic flux.



GRADIENT ECHOES AND 1D IMAGING

The Larmor equation

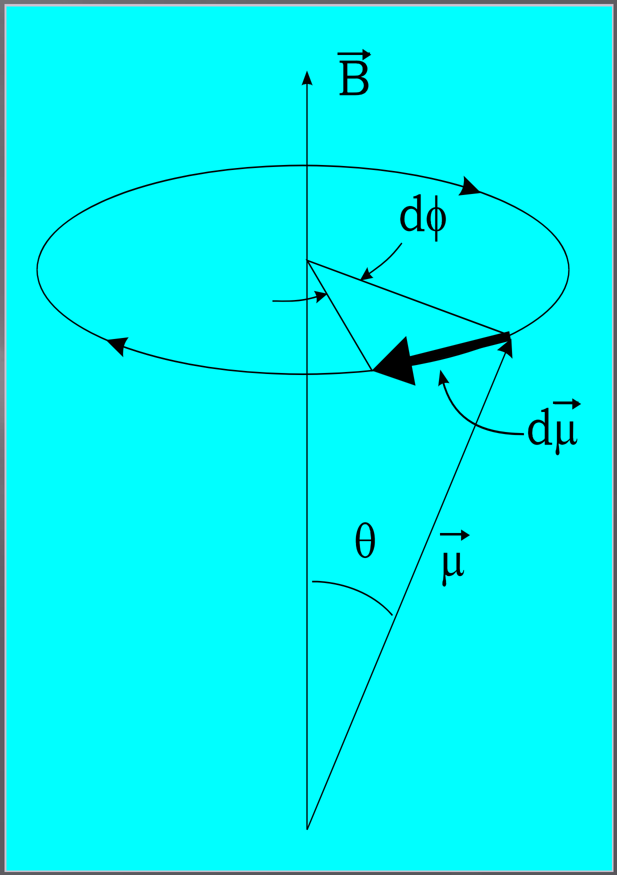
$$\boldsymbol{\omega} = -\gamma * \mathbf{B}$$

$$\phi = \omega t$$

$$\phi = -\gamma Bt$$

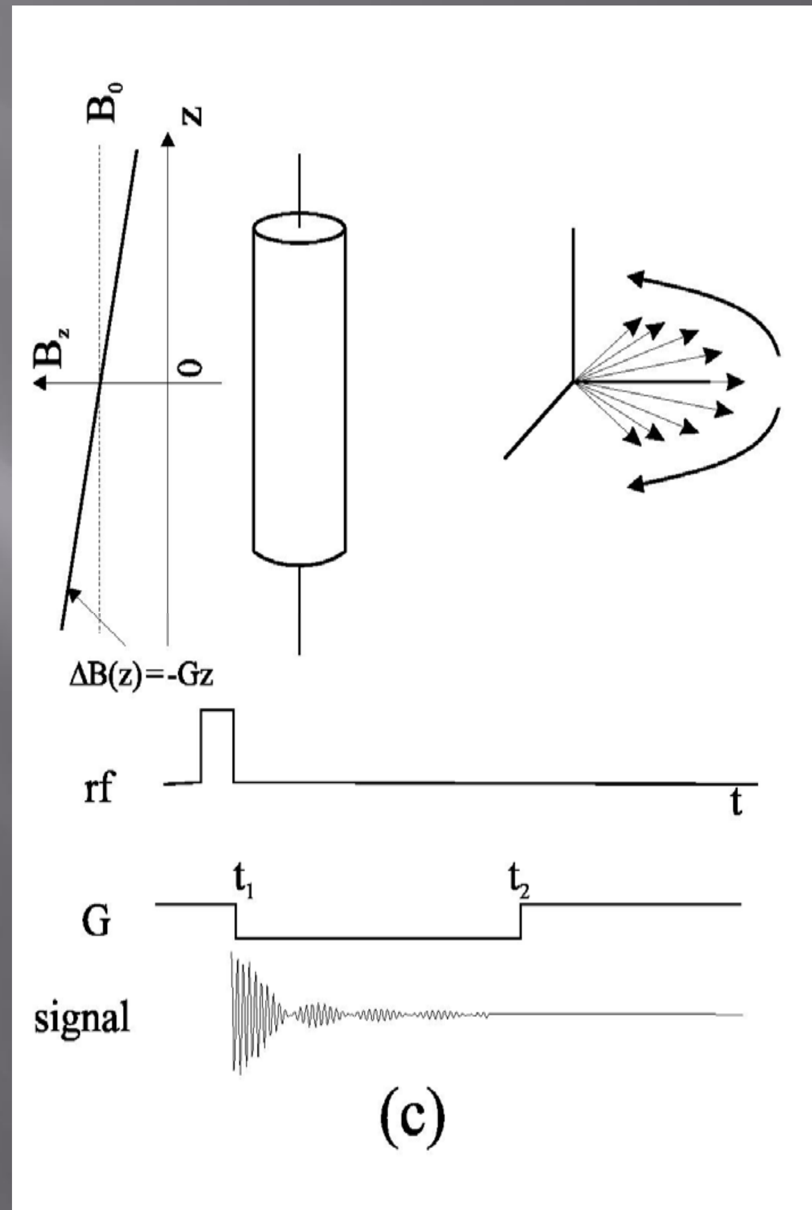
For hydrogen,

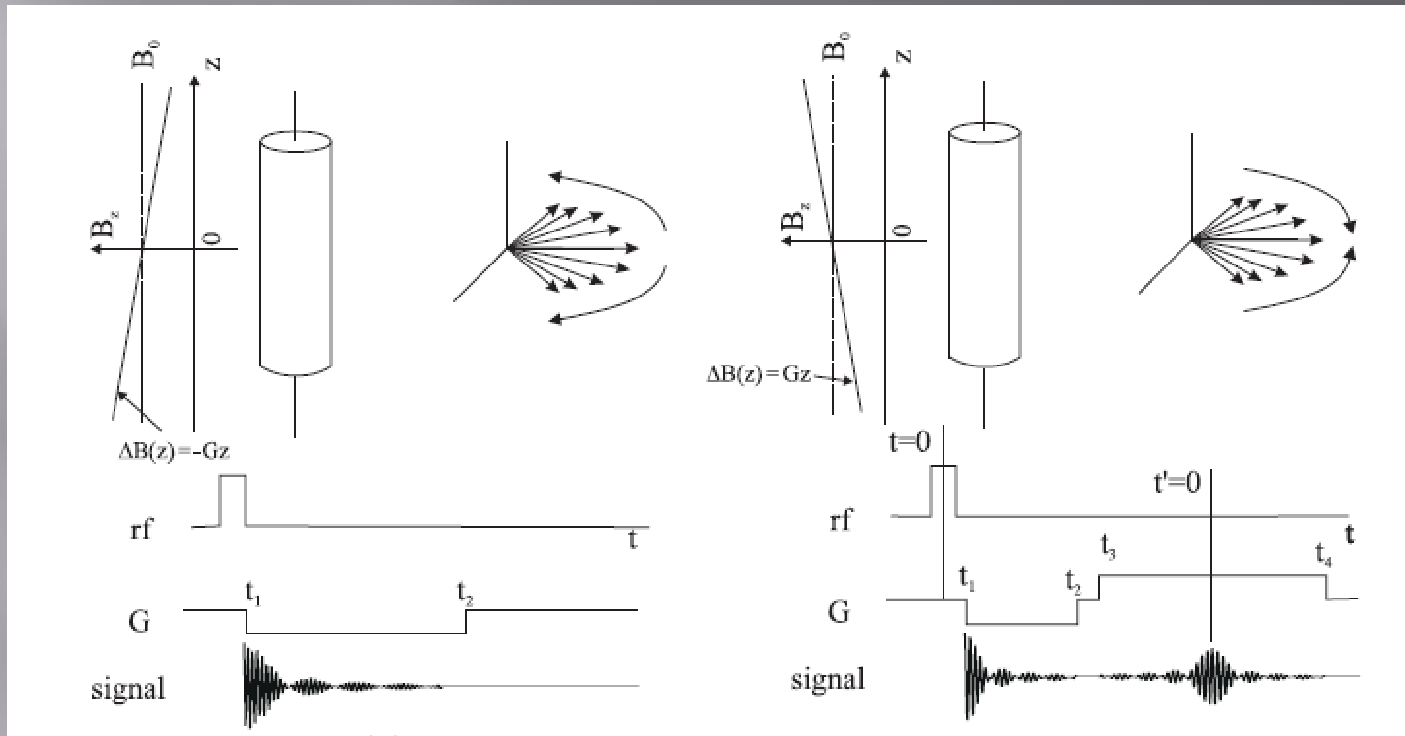
Ϋ = $\gamma / 2\pi = 42.6$ MHz/T.



By purposely adding a gradient field G, we can spatially distinguish the spins by their frequency content since now

$\omega(x) = \gamma\,(B_0 + Gx)$





Applying a dephasing gradient followed by a rephasing gradient causes the spins to first run away from each other and then to turn around and run back creating a gradient echo.

The MR signal as a Fourier transform

$$s(t) = \int dx \rho(x) \exp(-i\gamma G x t)$$

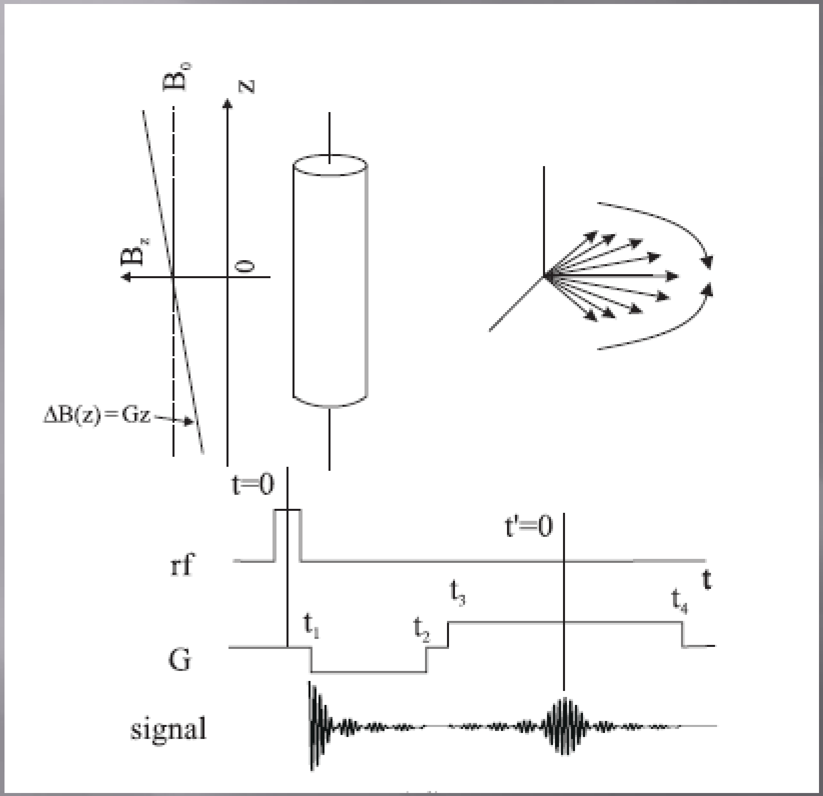
where ρ(x) is the spin density and

using k = ŸGt as the spatial frequency we find

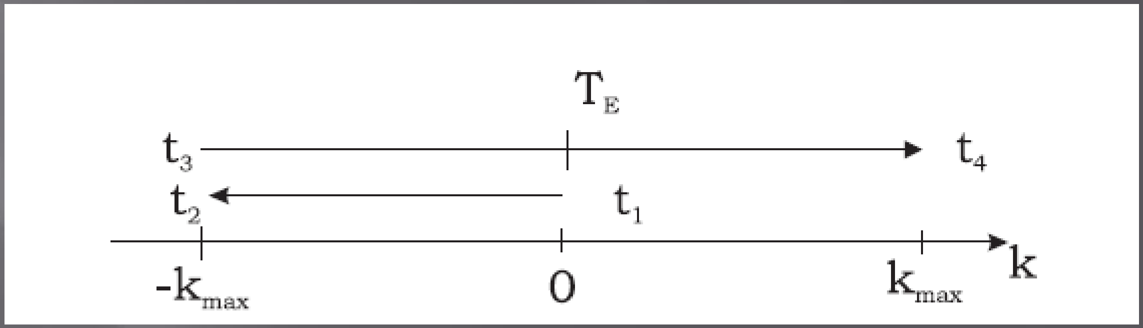
$$s(k) = \int dx \rho(x) \exp(-i 2\pi k x)$$

$$\rho(x) = \int dk\, s(k) \exp(i 2\pi k x)$$

Spatial encoding using a negative followed by a positive gradient to produce a gradient echo signal

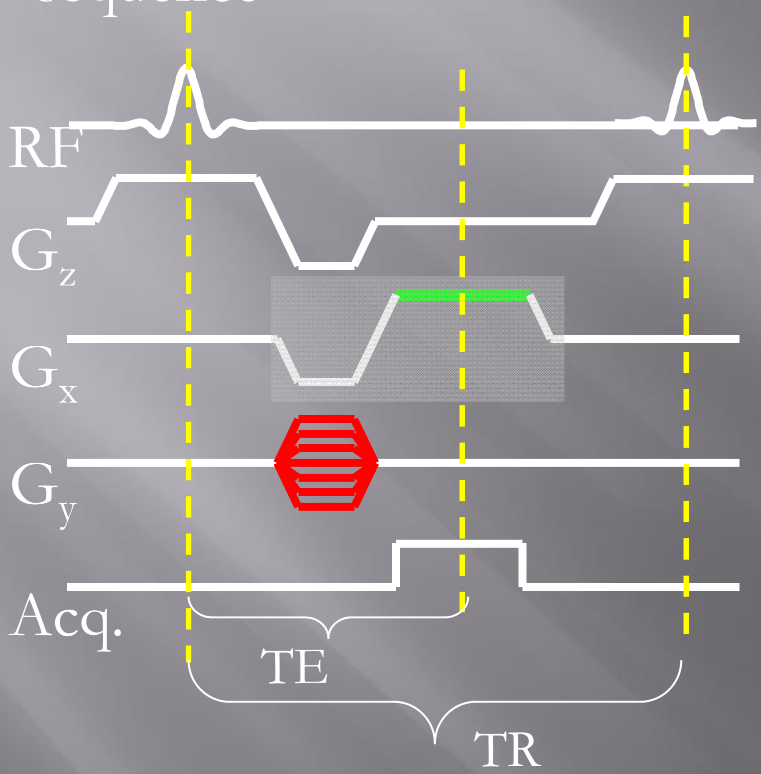


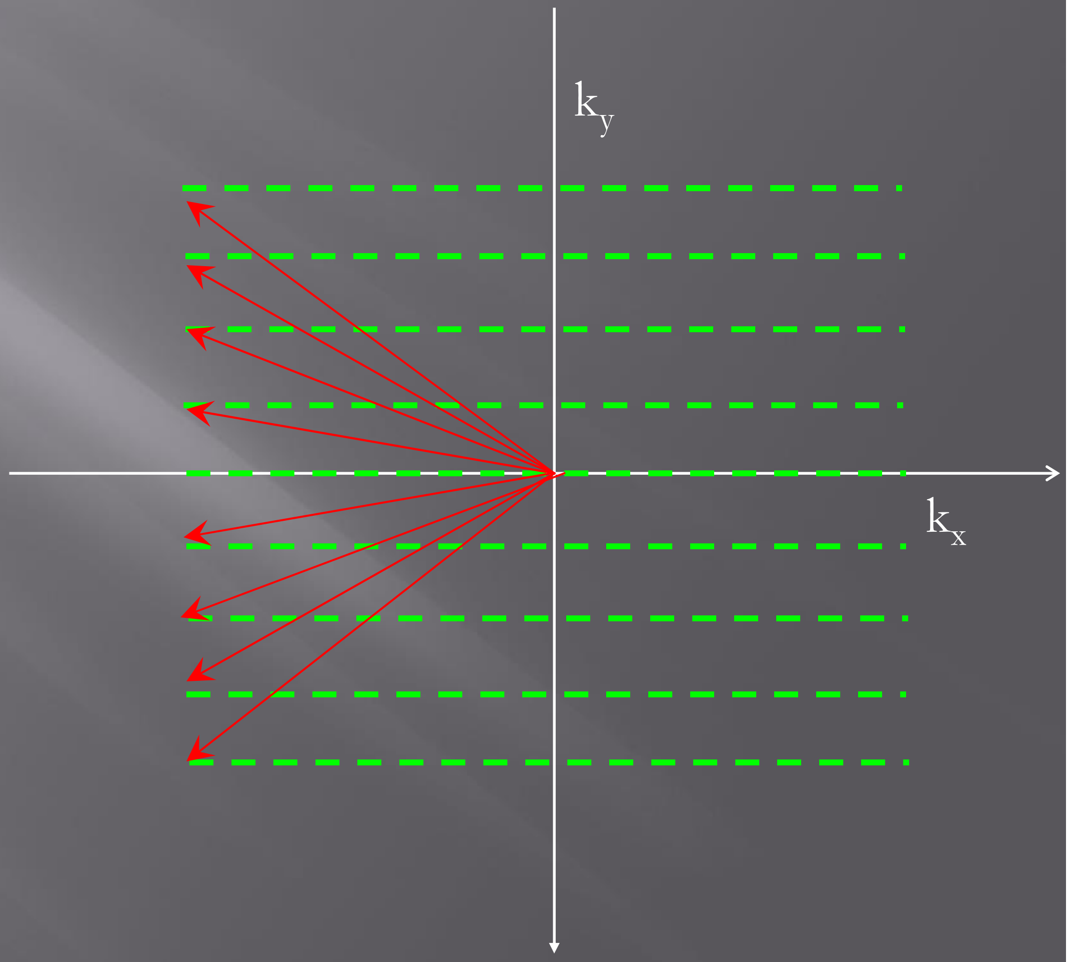
k-space



2D imaging: K-space trajectory

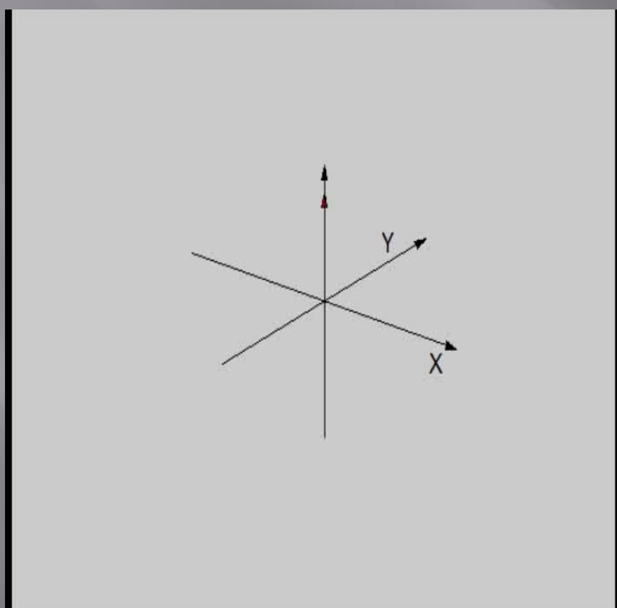
Typical Gradient Echo sequence





T1 relaxation

- Describes recovery of magnetization toward the equilibrium state



$$M_z(t) = M_o(1 - e^{-t/T1})$$

where t is the time between the rf pulse (with $\theta = \pi/2$) and the readout interval.

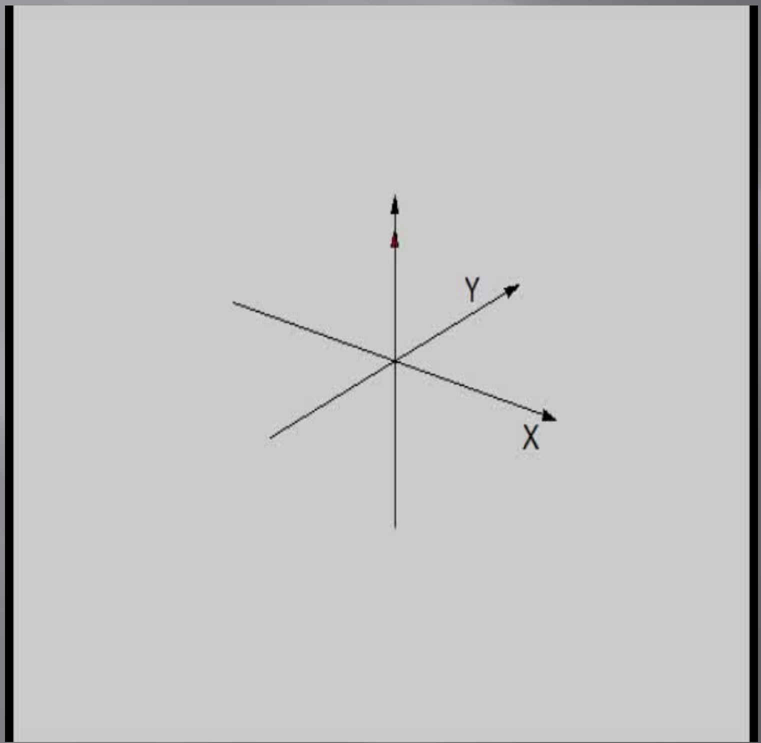
Longitudinal or T1 relaxation time

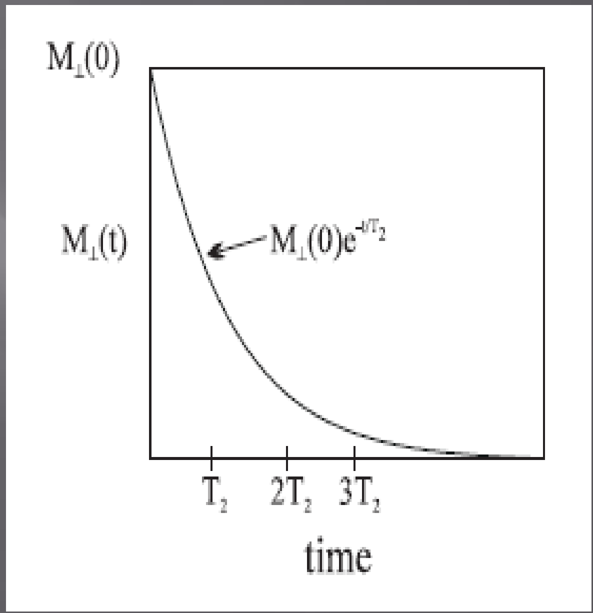
- Spin-lattice interaction
- Describes the regrowth rate of M_z toward the equilibrium value of M_o
- Definition: T1 is the time needed for M_z to recover from 0 to $(1-e^{-1})$ of the equilibrium value M_o

Tissue	T_1 (ms)	T_2 (ms)
gray matter (GM)	950	100
white matter (WM)	600	80
muscle	900	50
cerebrospinal fluid (CSF)	4500	2200
fat	250	60
blood[3]	1200	100-200[4]

T2 relaxation

- Free Induction Decay (FID)





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

T2 relaxation

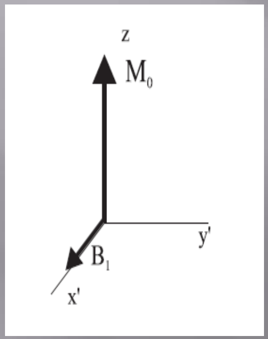
- Spin-Spin interaction
- Describes decaying rate of Mxy
- Practically, T2* decay, rather than T2 decay, is observed in FID due to field inhomogeneity

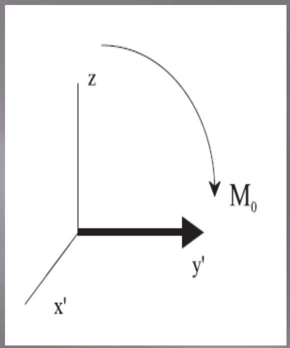
Tissue	T_1 (ms)	T_2 (ms)
gray matter (GM)	950	100
white matter (WM)	600	80
muscle	900	50
cerebrospinal fluid (CSF)	4500	2200
fat	250	60
blood[3]	1200	100-200[4]

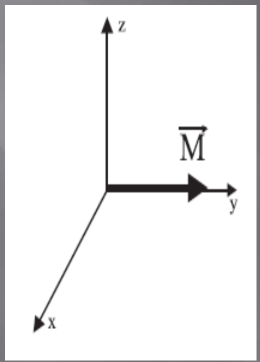
T2* relaxation

- $$\frac{1}{T_2*} = \frac{1}{T_2} + \frac{1}{T_2'} > \frac{1}{T_2}$$
- Combined external field inhomogeneity effects (T2') and thermodynamic effects (T2)
- T2' origins:
  - External: B_0 field, boundaries, partial volume effect
  - Internal : chemical shift, microscopic motion (diffusion, capillary blood)
  - Effective spatial scale: ~voxel size to molecular size

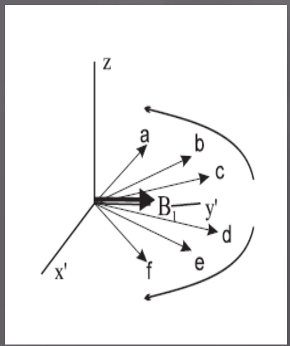
T2* relaxation







T2 decay (ideal)



T2* decay (actual)

T2* Weighted Contrast

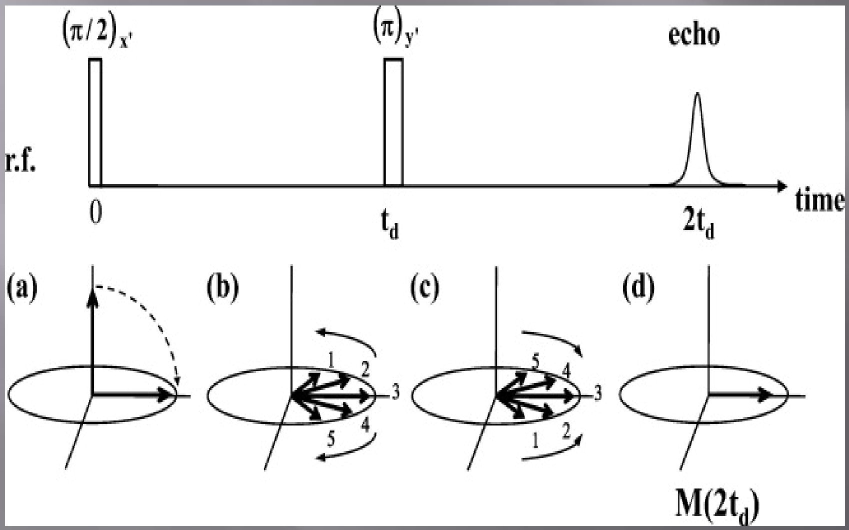
- Directly collect the FID using long echo time TE:

$$M_{xy}(TE) = M_{xy}(0)e^{-\frac{TE}{T2*}}$$

- Typical T2*W sequence:
  - Gradient Refocused Echo (GRE)
  - Echo Planar Imaging (EPI)

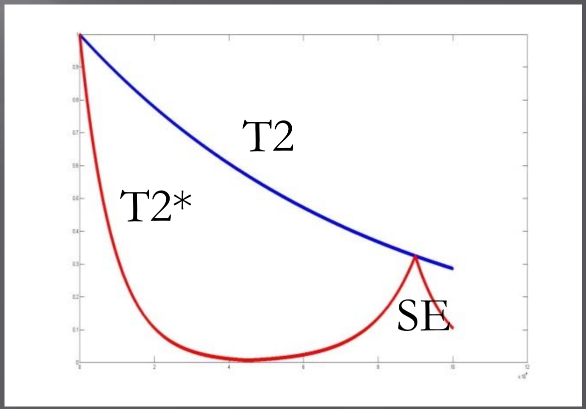
T2 Weighted Contrast

- Dephase from field inhomogeneity (or background gradient) can be refocused by a π rf pulse



$$M_{xy}(TE) = M_{xy}(0)e^{-\frac{TE}{T2}}$$

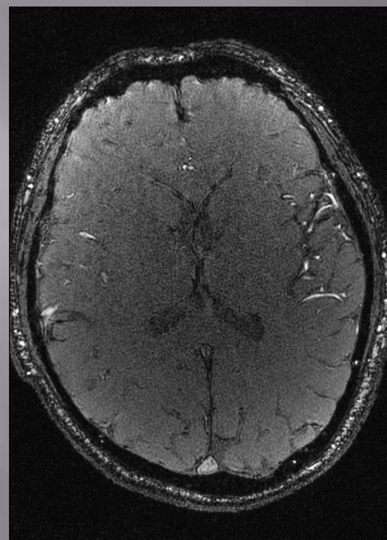
(valid at TE=$2t_d$)



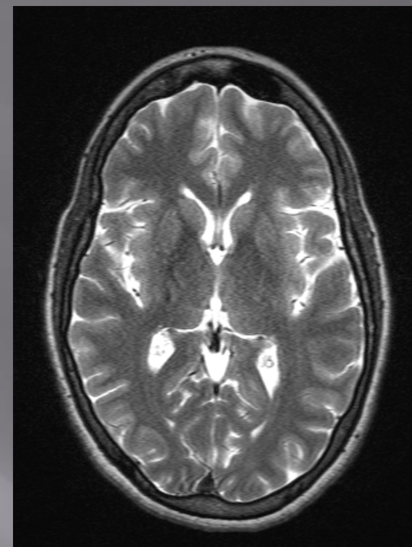
- T2W sequence:
  - Spin Echo (SE)
  - Fast Spin Echo (FSE)

GRE T1

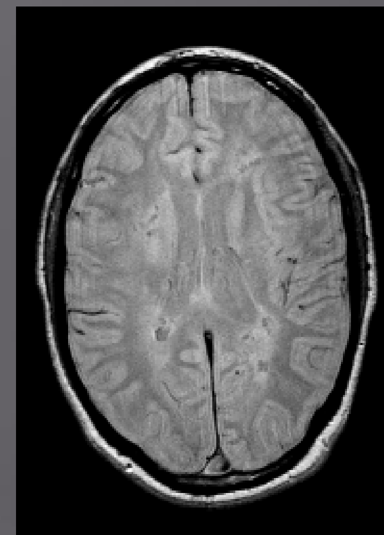
GRE T2*

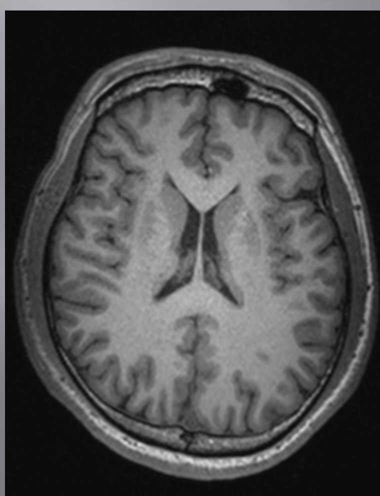


SE T2



GRE PD

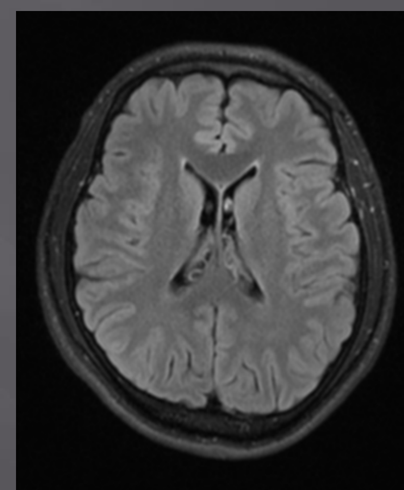




IR T1 (MPRAGE)

IR T1, WM null

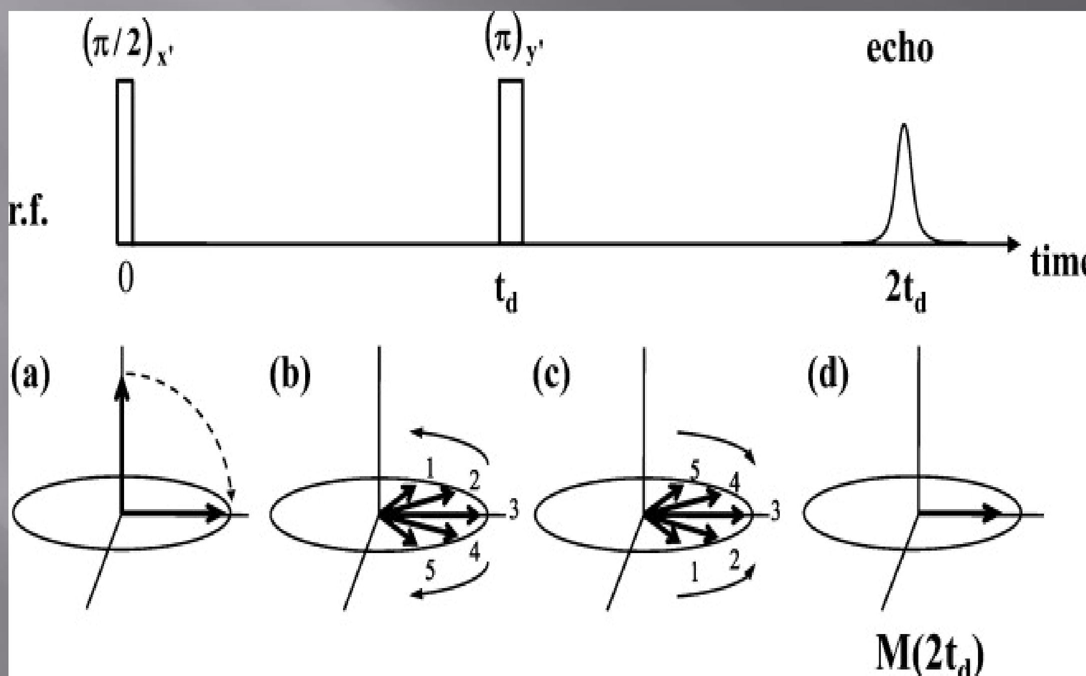
IR T1, GM null



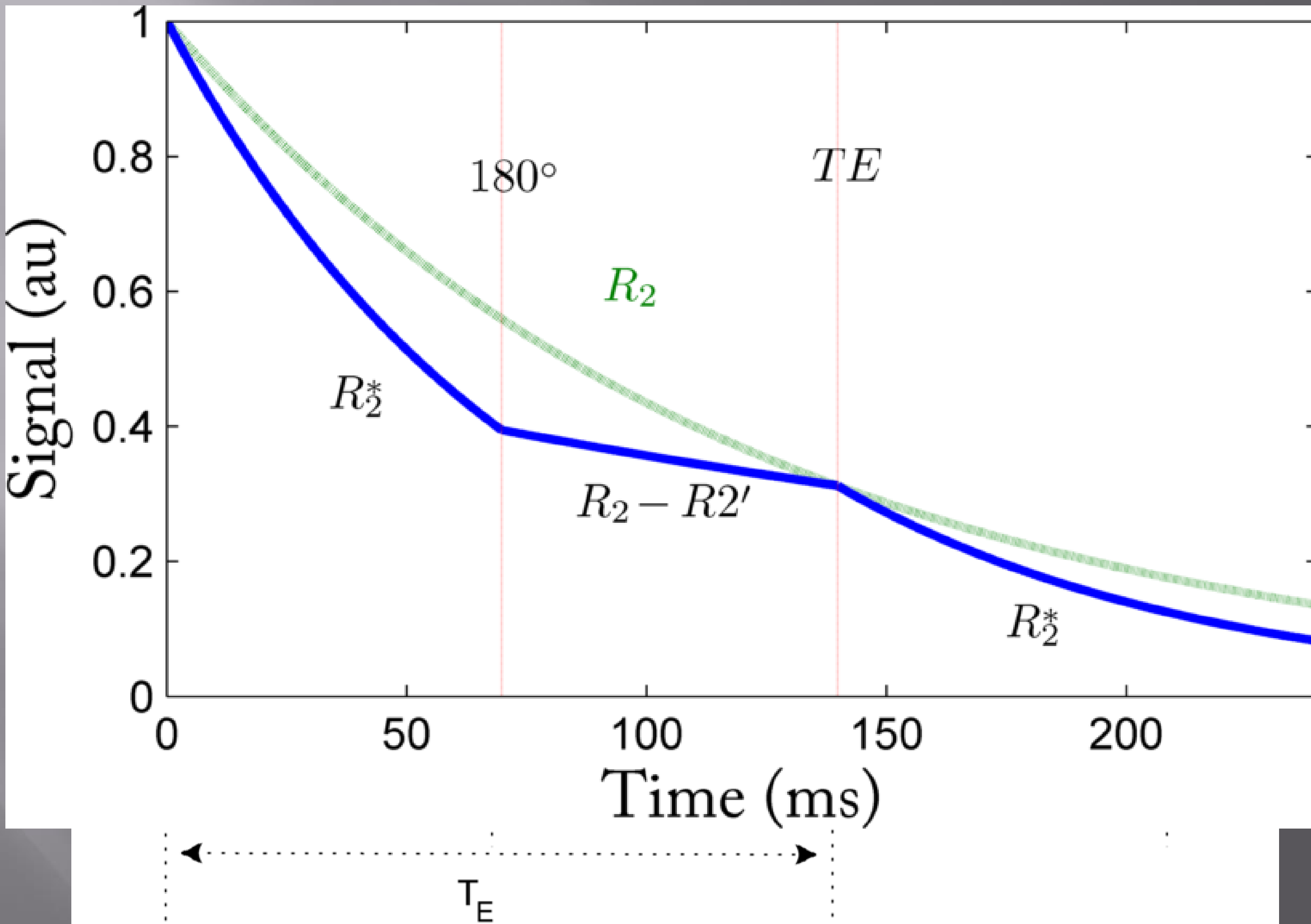
T2, IR CSF null
(FLAIR)

GESFIDE SEQUENCE

Wei Feng, PhD

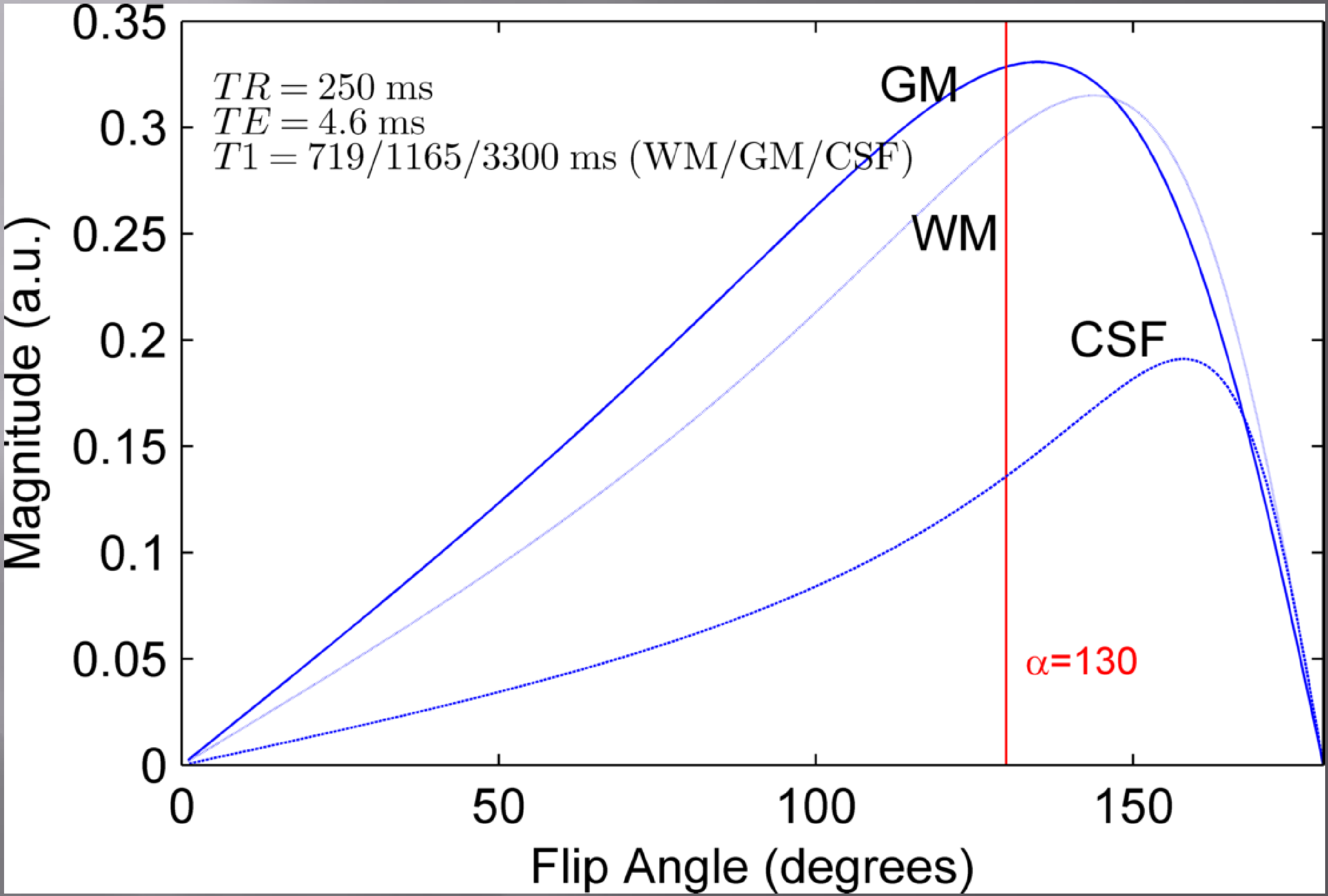


The GESFIDE Sequence[1]



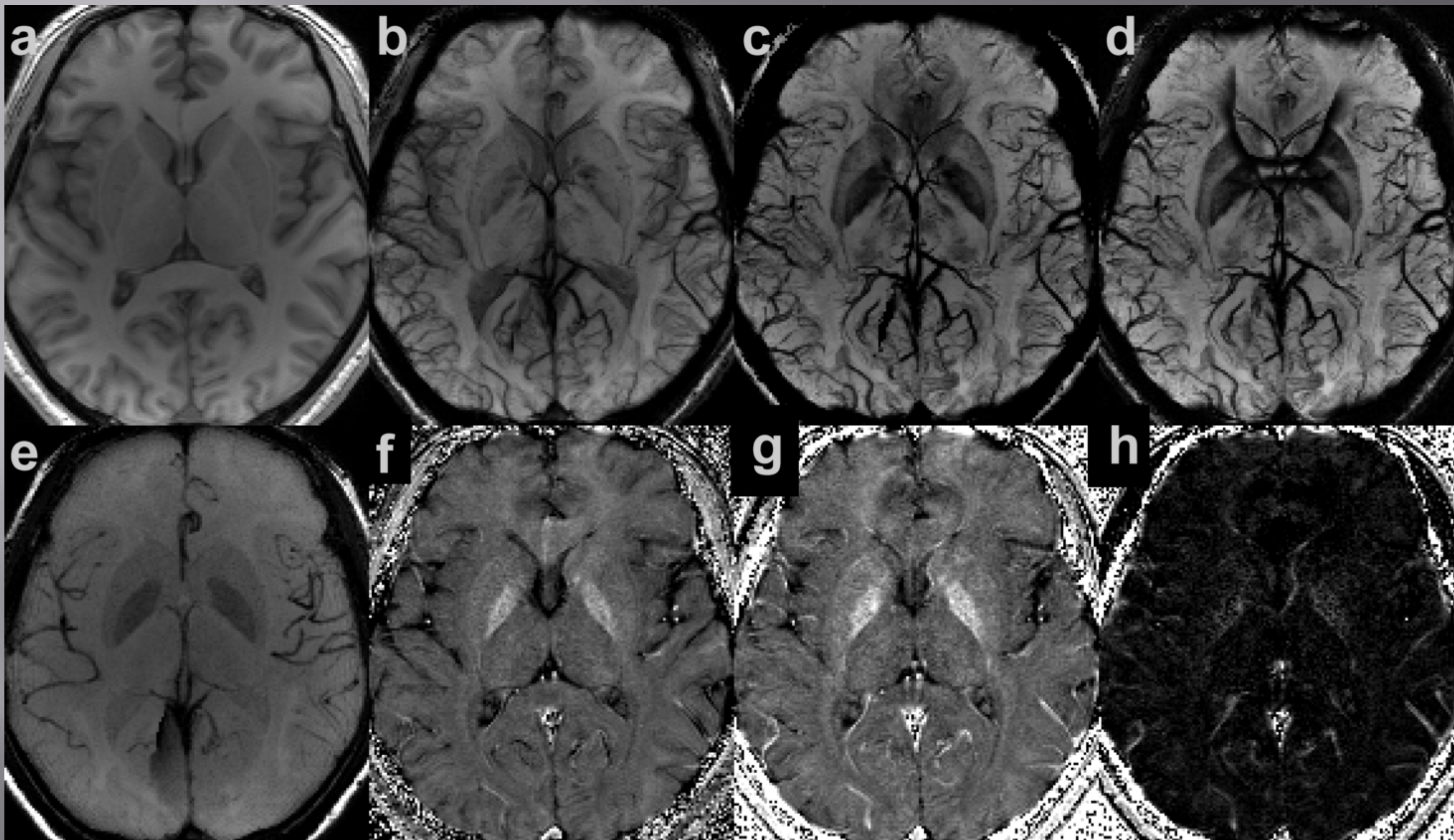
[1.] Ma and Wehrli, 1996; Yablonski and Haacke, 1997

Ernst Angle Optimization



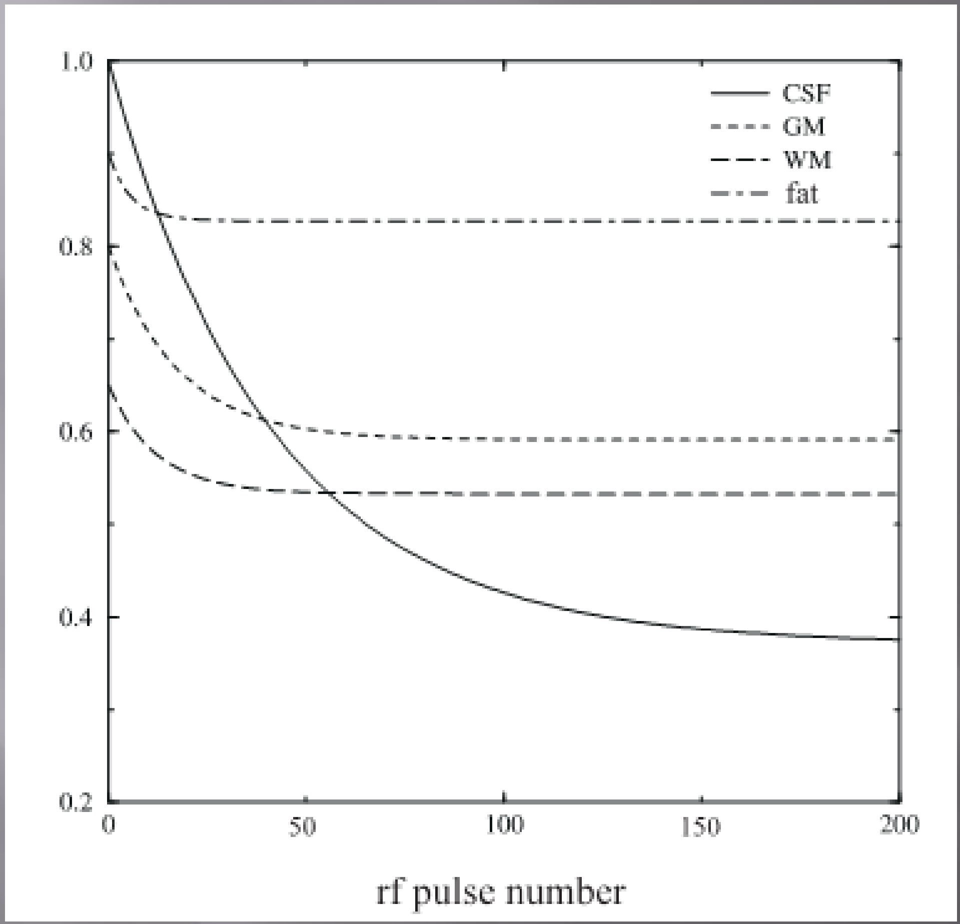
Signal magnitude at the first echo as a function of the flip angle.

Imaging Results



(a) T1W image at first echo; (b) SWI image at TE=21.04ms; (c) SWI image at TE=79.66ms (averaged over 5 echoes from 74.18ms to 85.14ms; (d) Same as (c) but generated with conventional multiecho SWI; (e) MRA at spin echo; (f), (g) and (h): R2, R2* and R2' maps. Note (b), (c), (d) and (e) were generated with minimum intensity projection over 8 slices whose center slice was aligned with the slices shown in (a), (f), (g) and (h).

Approach to Steady State



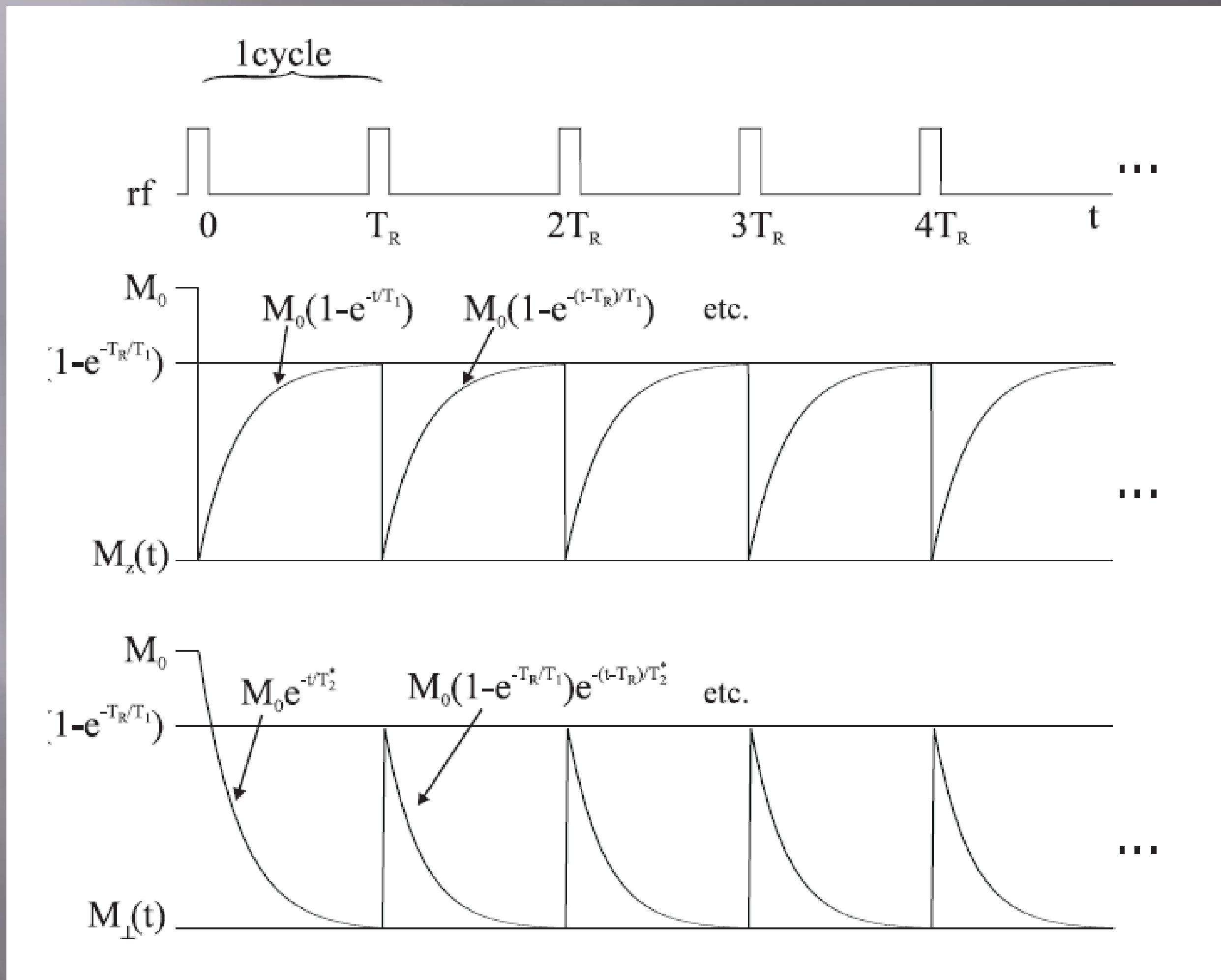
$$\theta = 10^\circ$$

$$T_R = 40\,\mathrm{ms}$$

$$\hat{\rho}(\theta, T_E) = \rho_0 \sin\theta \frac{(1 - E_1)}{(1 - E_1 \cos\theta)} e^{-T_E/T_2^*}$$

$$E_1 \equiv e^{-T_R/T_1}$$

General signal behavior



ρ*(1-E1)*E2* where E1 = exp(-TR/T1) and E2 = exp(-TE/T2*)

Why choose 90^o flip angle?

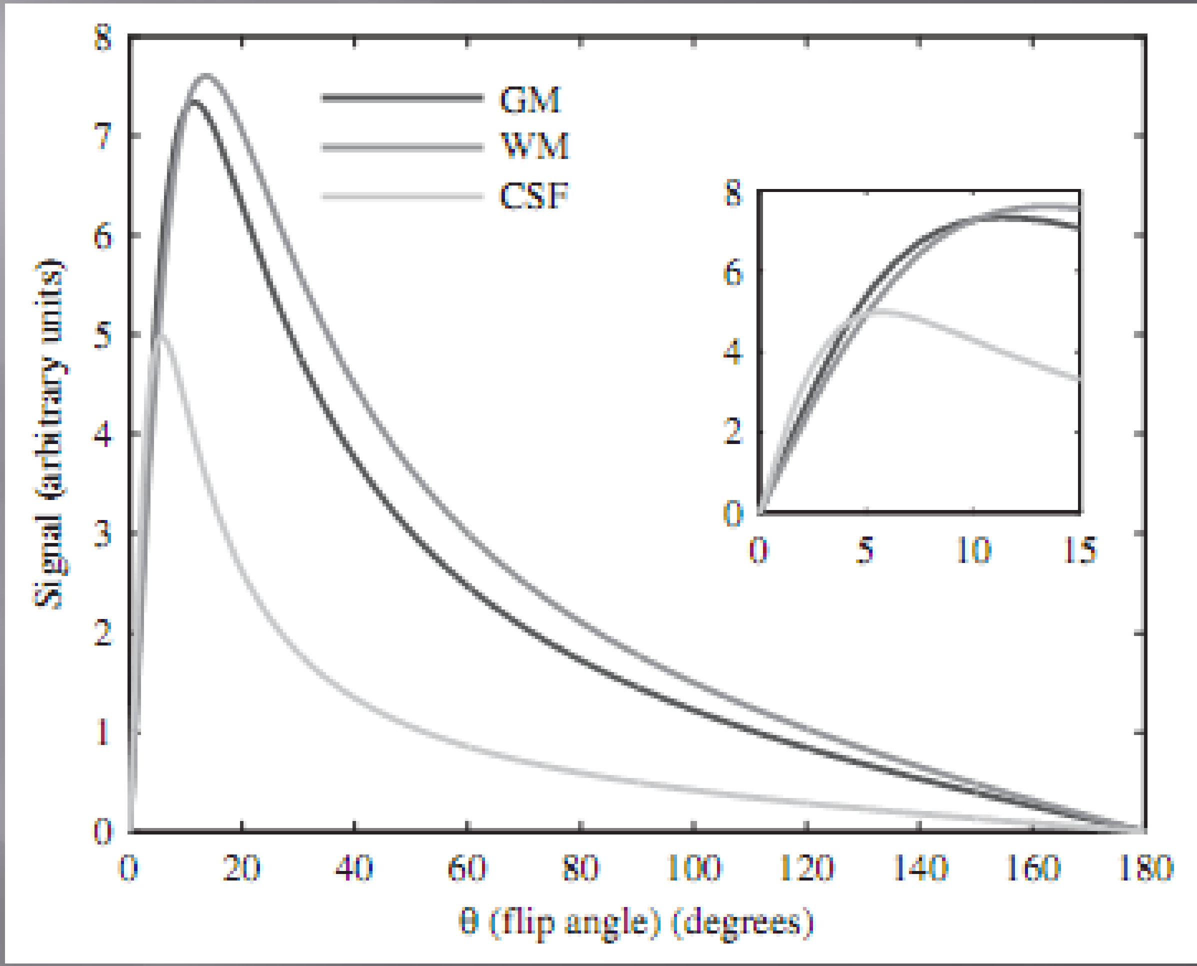
It limits the total T1 recovery and for small TR the signal only recovers to TR/T1. If TR = 10ms and T1 = 2000ms, the signal is only 0.5% of its maximum value.

Now if we use a smaller flip angle, then we find the signal becomes:

$$\hat{\rho}(\theta, T_E) = \rho_0 \sin\theta \frac{(1 - E_1)}{(1 - E_1 \cos\theta)} e^{-T_E/T_2^*}$$

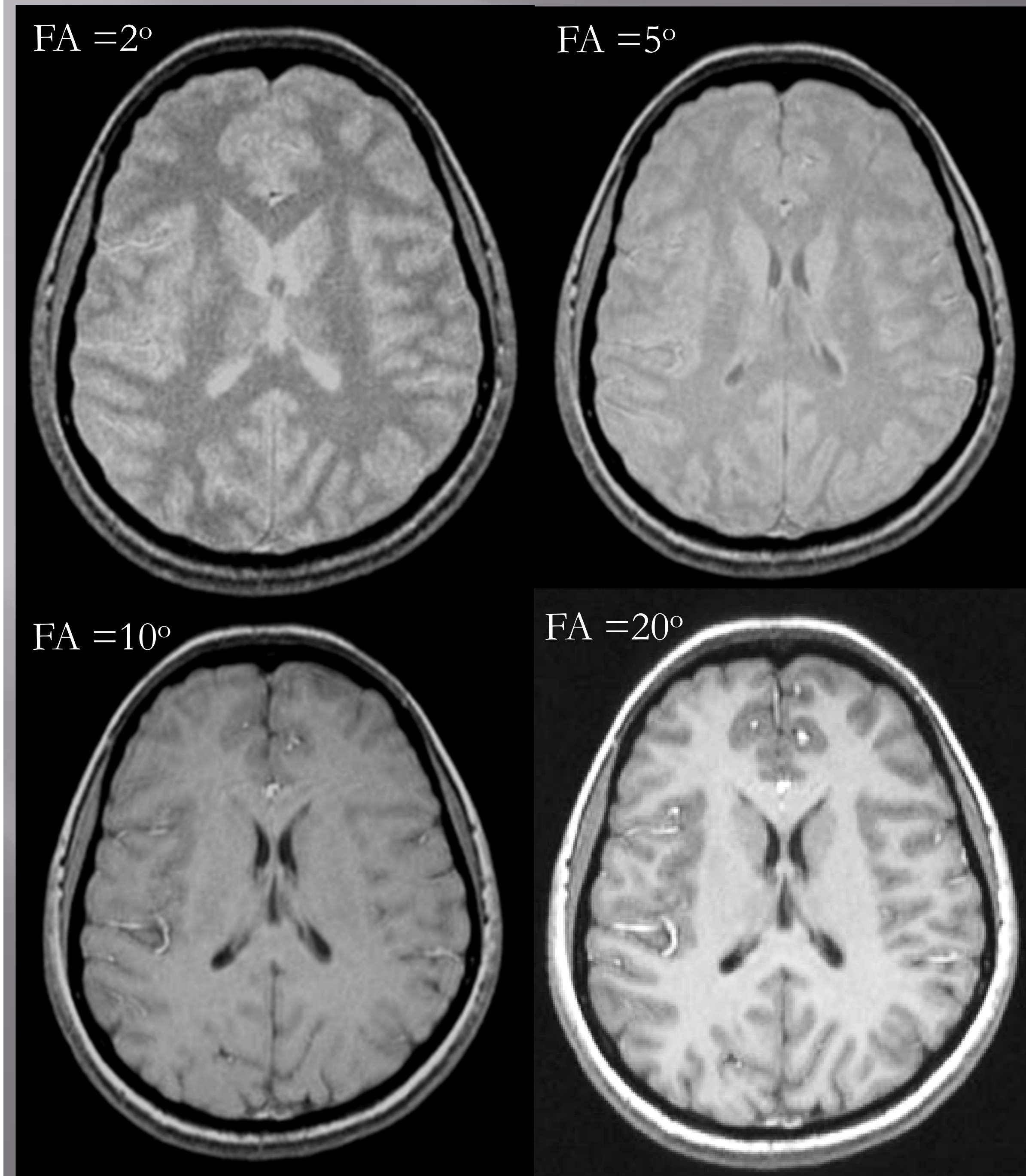
At the optimal flip angle (also known as the Ernst angle), given by $\cos\theta = E1$, we find the peak signal is roughly 0.5sqrt(2TR/T1) or 5% of the total (that is 10 times bigger).

Signal behavior at 1.5T for a TR of 20msec



$$\hat{\rho}(\theta, T_E) = \rho_0 \sin\theta \frac{(1 - E_1)}{(1 - E_1 \cos\theta)} e^{-T_E/T_2^*}$$

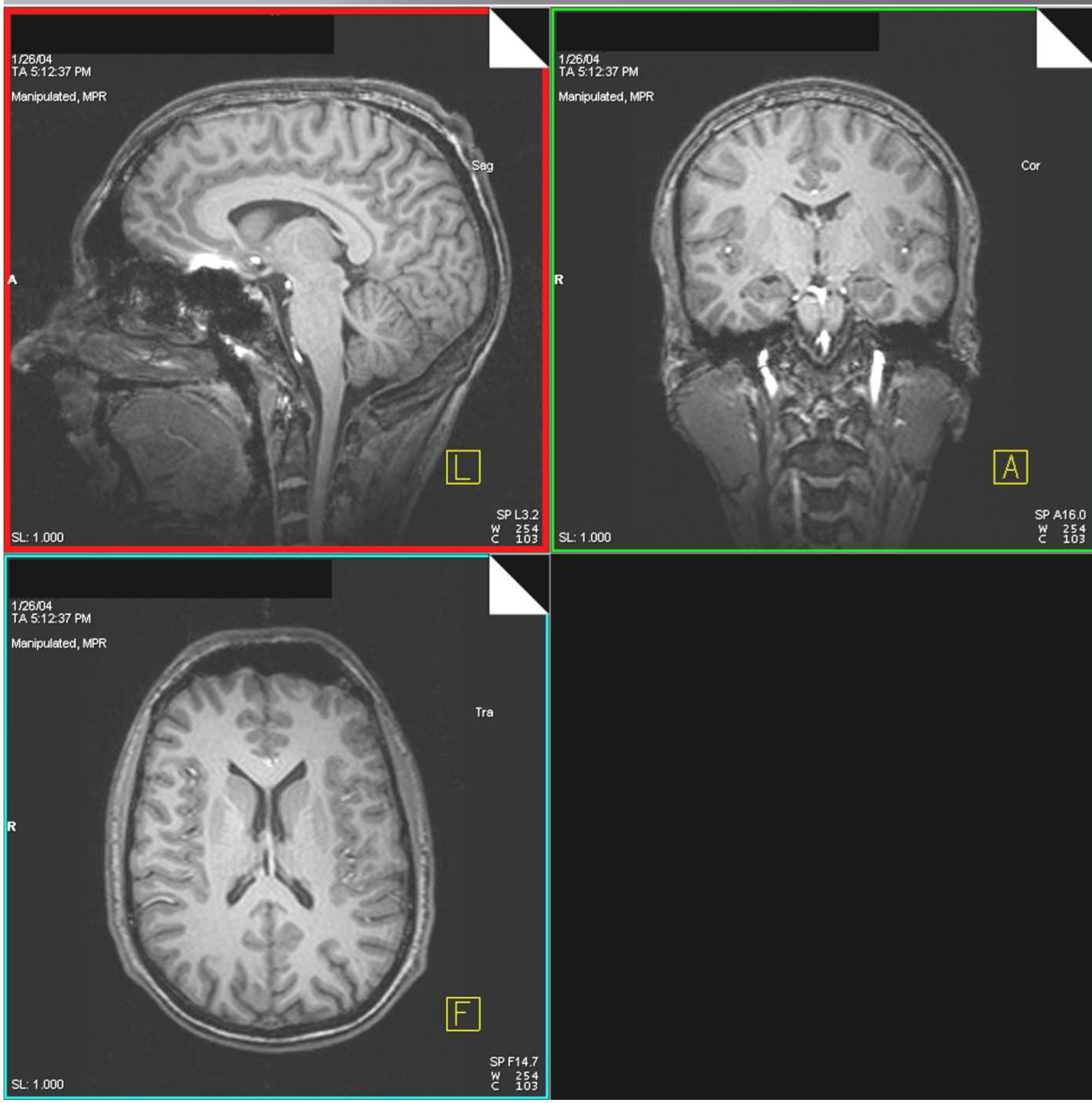
$$E_1 \equiv e^{-T_R/T_1}$$



Example data from a 3D gradient echo sequence with short TE and short TR = 20ms at 1.5T.

Small flip angles generate a spin density weighted image while large flip angles generate a T1 weighted image.

3D MP-RAGE

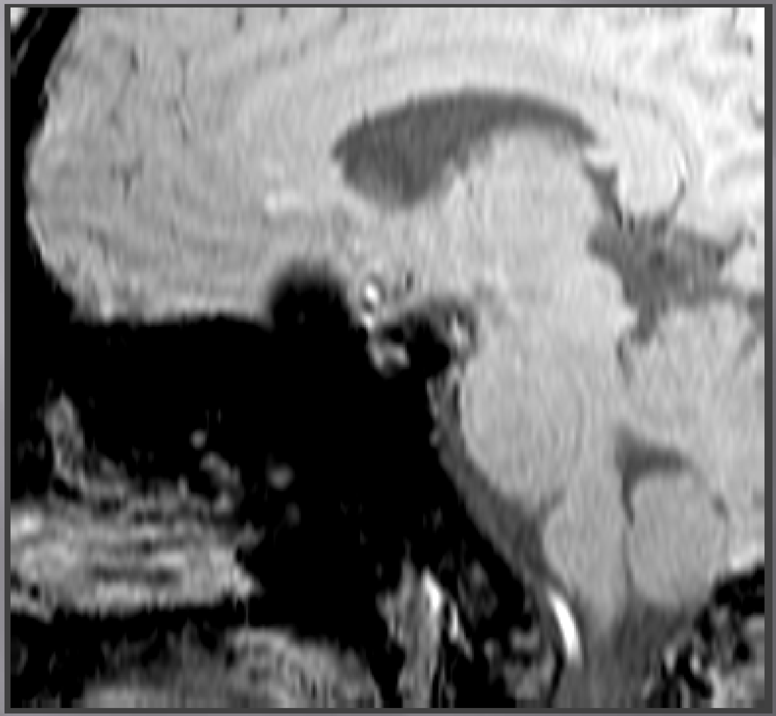


Structural Imaging:

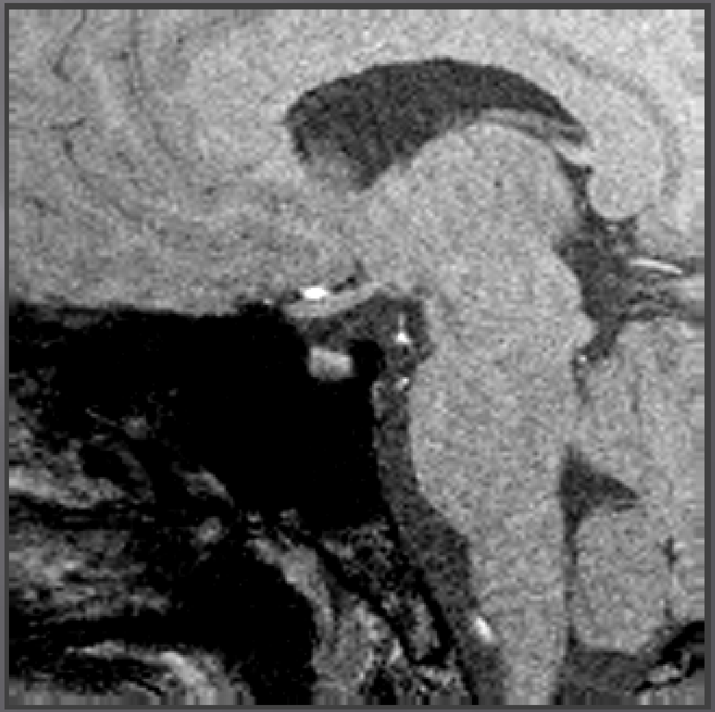
- T1 3D MPRAGE
- 1.0 x 1.0 mm² in plane resolution
- 1 mm slice thickness
- 176 slices
- TE 5 ms
- TR(total) 2500 ms
- Flip angle: 12 °
- Bandwidth 200 Hz/Px
- Scan time: 10:32 min

Effect of Bandwidth on MR image quality

Bandwidth 110 Hz/Px

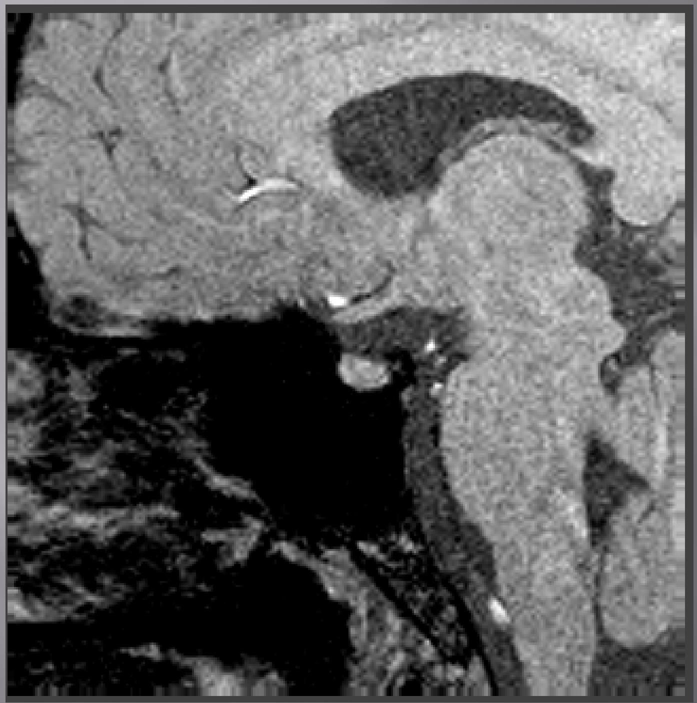


Bandwidth 473 Hz/Px



Local field distortions at short TEs

TE=10ms



0.7x0.7x1.4mm^3

TR=14ms, Flip angle:12°

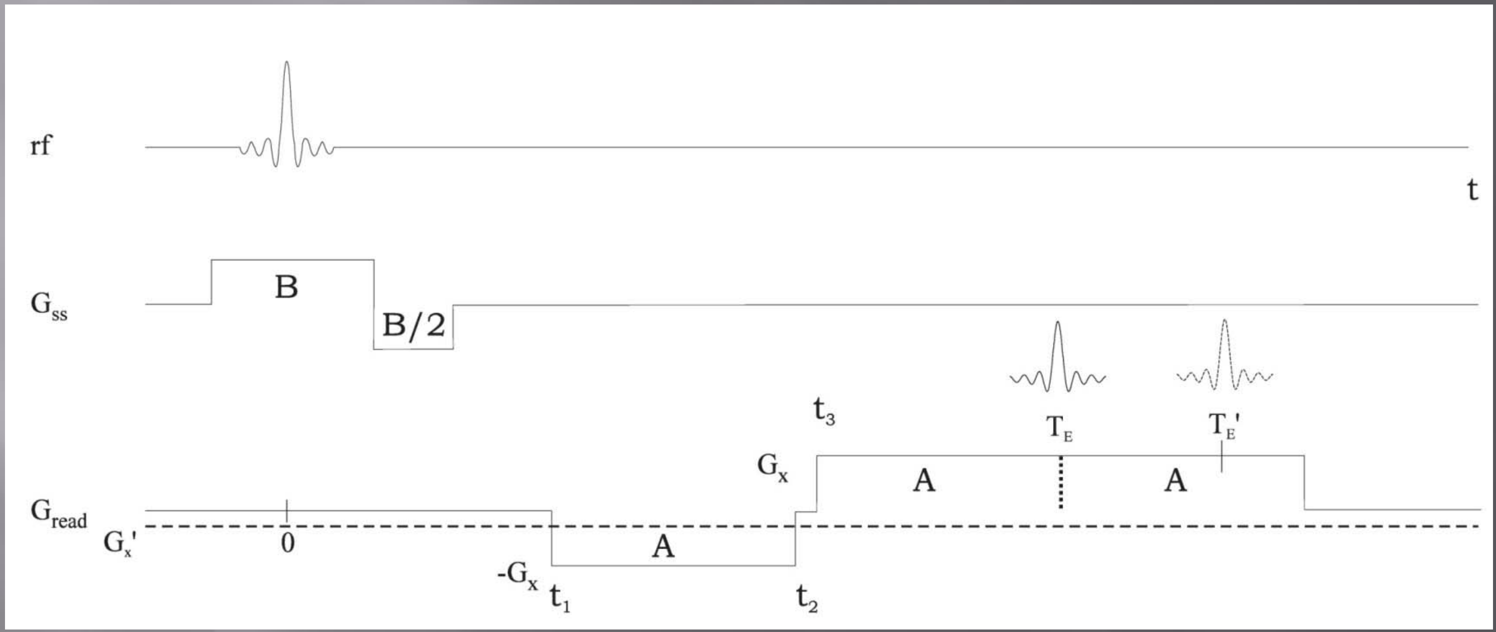
Bandwidth=473 Hz/Px

TE=2.5ms

0.7x0.7x1.4mm^3

TR=12ms, Flip angle:12°

Bandwidth=473 Hz/Px



Sequence diagram for double-echo gradient echo sequence. Haacke, Brown, Thompson, Venketesan Wiley, 1999, page 583
Magnetic Resonance Imaging: Physical Principles and Sequence Design

How to choose the right resolution and echo time

$$N_{shift}\Delta t = [G'/(G + G')]TE$$

$$\Delta x_{nsl} = 12/(\delta B_0 TE)\text{mm}$$

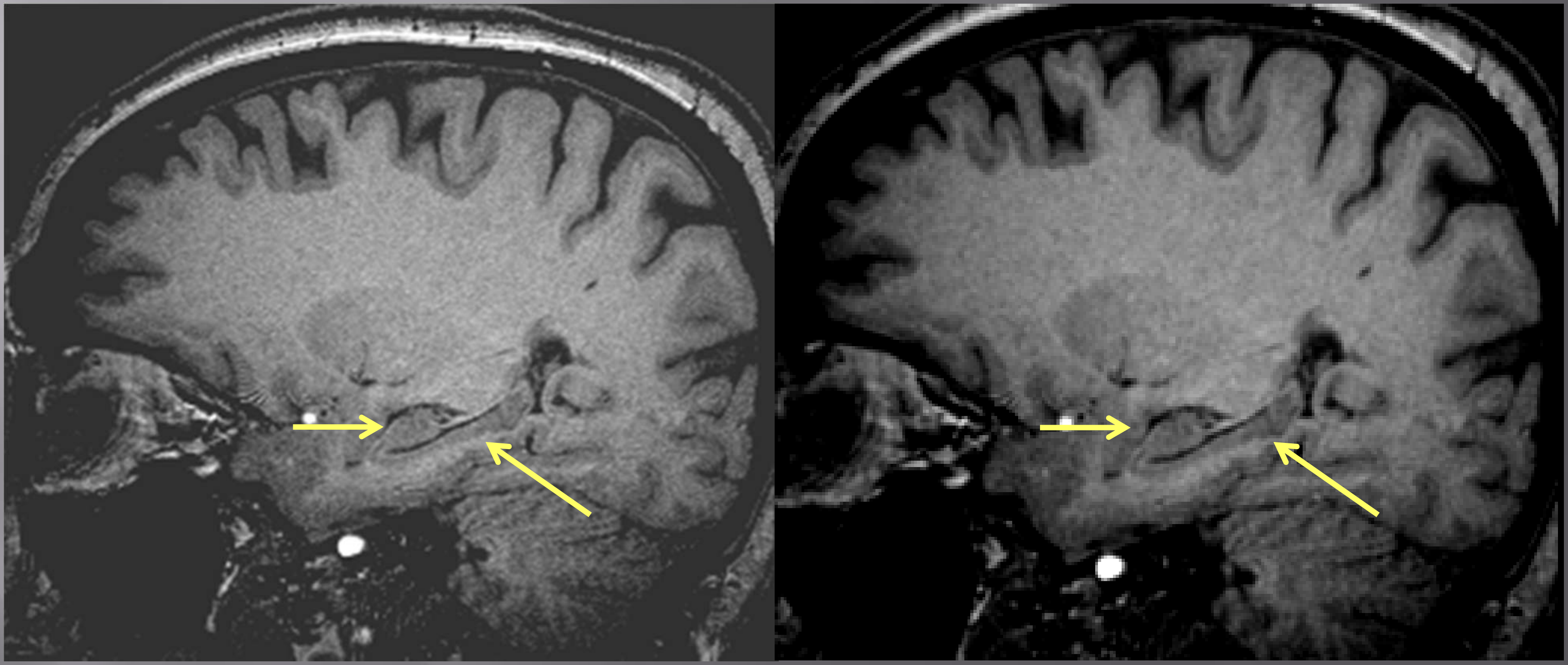
with TE in ms and δ is in ppm/mm

nsl = no signal loss

Sagittal images: 0.6mm x 0.6mm x 1.2mm

Original image

Sliding window filtered



Note this is not the happy campers structure but rather the

Hippocampus

MRA

resolution:
0.5mm x
0.5mm x
1.0mm

1mm resolution vertically

0.25mm resolution vertically

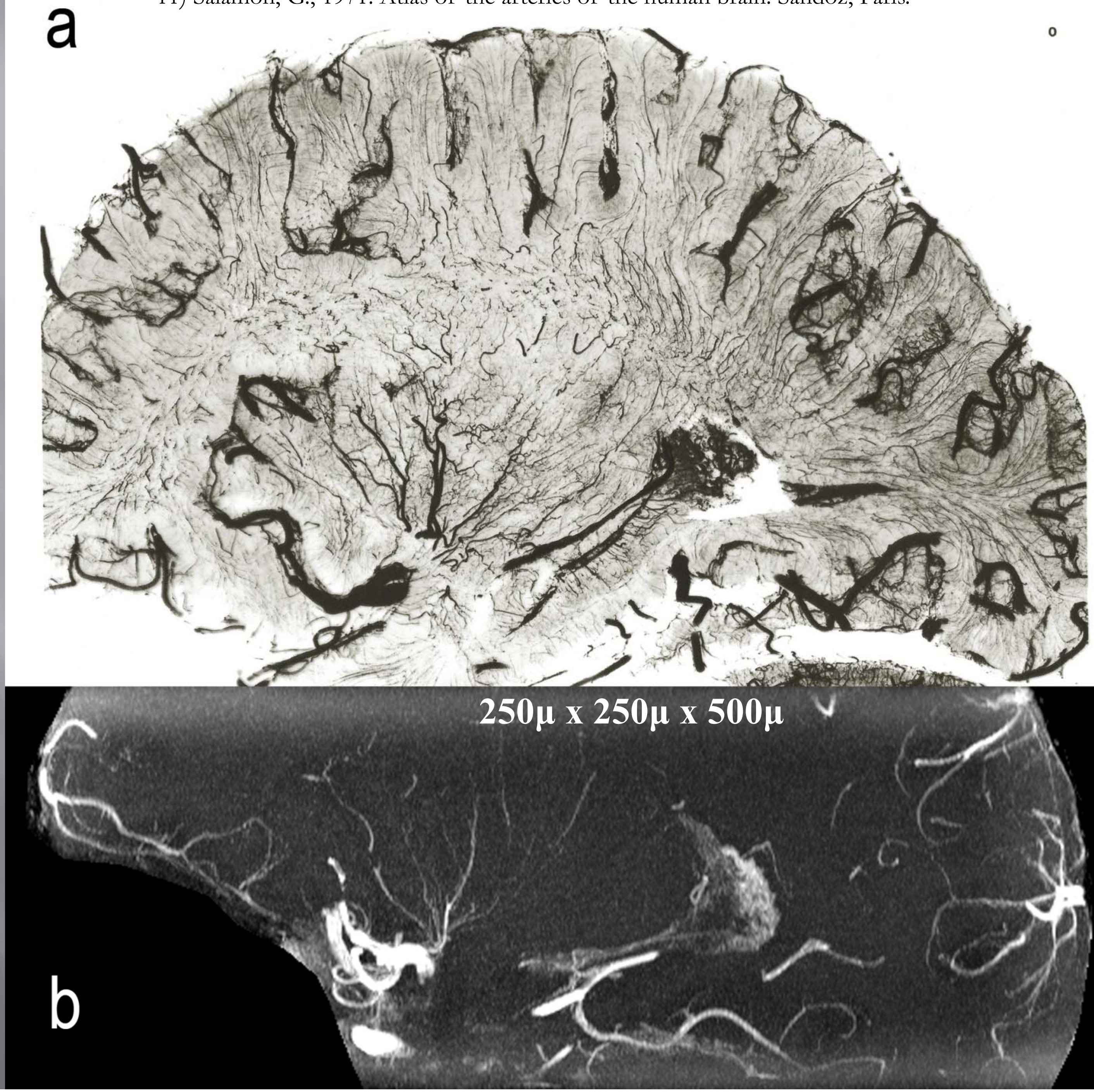
High resolution MR angiography

Small arteries around 250 microns are beginning to become visible even without a contrast agent (0.5mm isotropic resolution).

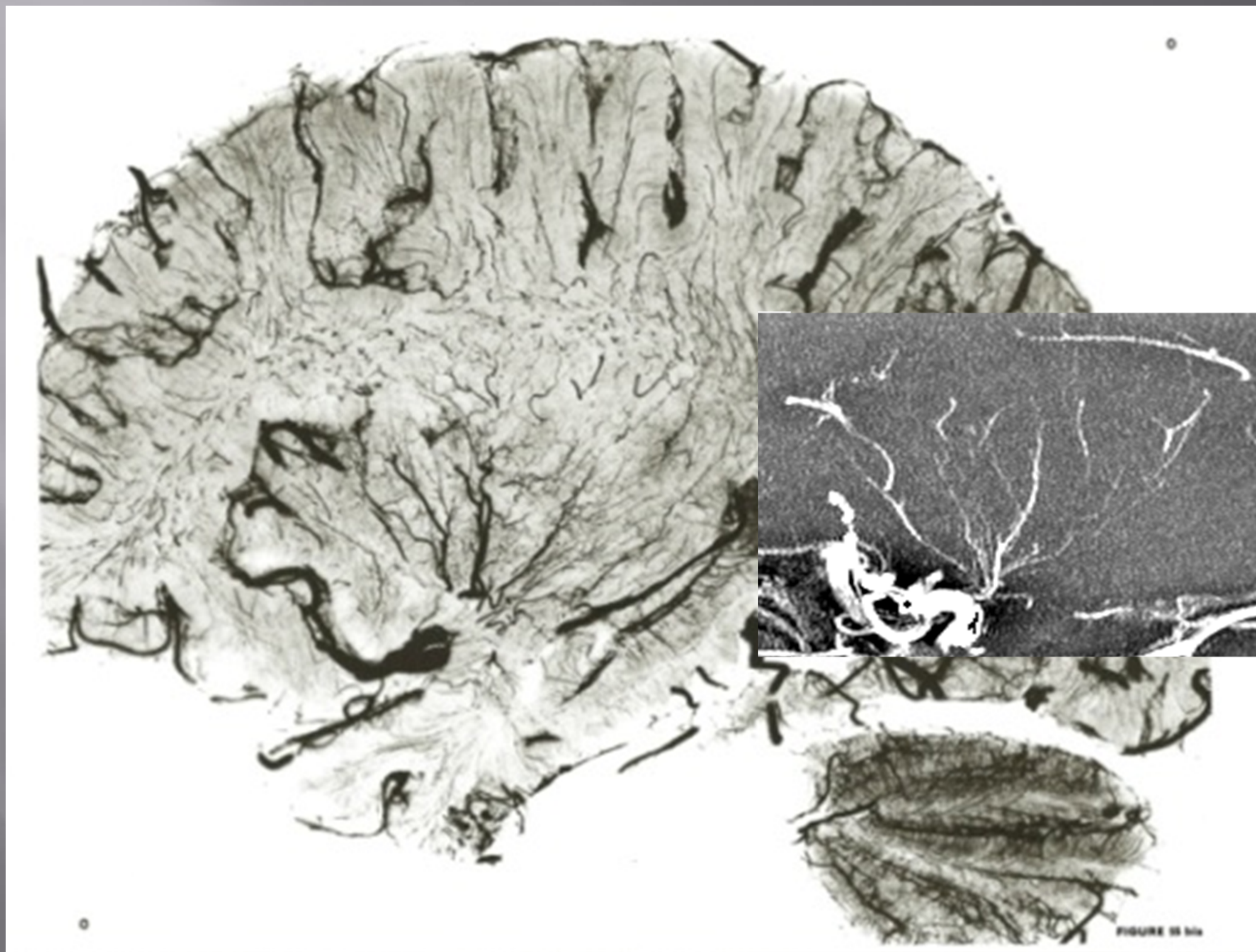
Cadaver brain imaging of arteries using an injection technique

Salamon, G., 1971. Atlas of the arteries of the human brain.

11) Salamon, G., 1971. Atlas of the arteries of the human brain. Sandoz, Paris.



Thalamic arteries



Conclusions

- Gradient echo imaging can be used for a variety of contrast types
- It is fast with good SNR and high resolution
- It can be used for SWI and MRA
- It is perhaps the most robust method to use from field strength to field strength and manufacturer to manufacturer