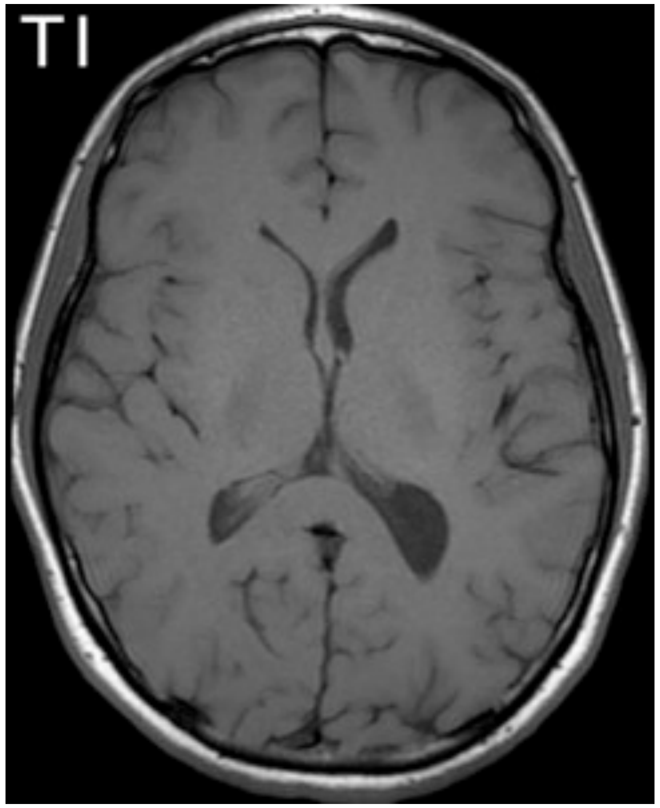
Introduction to Magnetic Resonance Imaging (MRI)

Alexandre Gramfort
alexandre.gramfort@telecom-paristech.fr

11 Oct 2012

TELECOM ParisTech

MRI



Magnetic Resonance Imaging (MRI) is a non-invasive imaging modality based on "Nuclear Magnetic Resonance"

- 3D
- High resolution (1mm isotropic for anatomy)
- Anatomical images, functional images, angiography, diffusion
- Many applications: clinical, cognitive neurosciences

Measures the magnetic properties of material

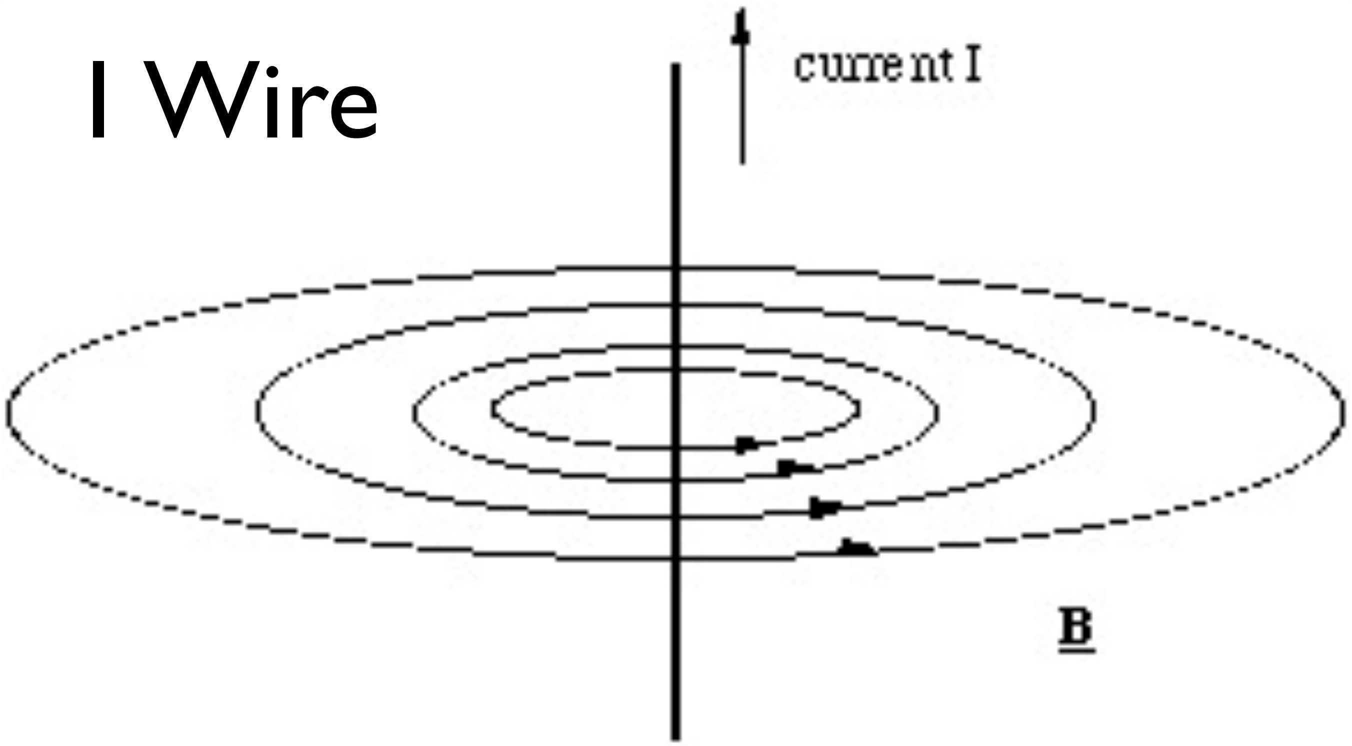
History

- 1945 : Bloch and Purcell "Nuclear induction" or "Nuclear magnetic resonance" (Nobel 1952)
- 1972 : 1^{st} scanner shows that tumor tissue is different from normal tissue in vivo with NMR (Damadian)
- 1973 : First image of a finger by Lauterbur (Nobel 2003 with Mansfield)
- 1975 : Imaging with Frequency and Phase encoding by Ernst (Nobel 1991)
- 1978 : First clinical systems (1982 in France)
- 1985 : Diffusion MRI (Le Bihan)
- 1990 : BOLD (Blood-oxygen-level-dependent) effect in vivo (Ogawa et al.)
- 1992 : BOLD activation experiments (Kwong et al. & Kwong et al. & Bandettini et al.)

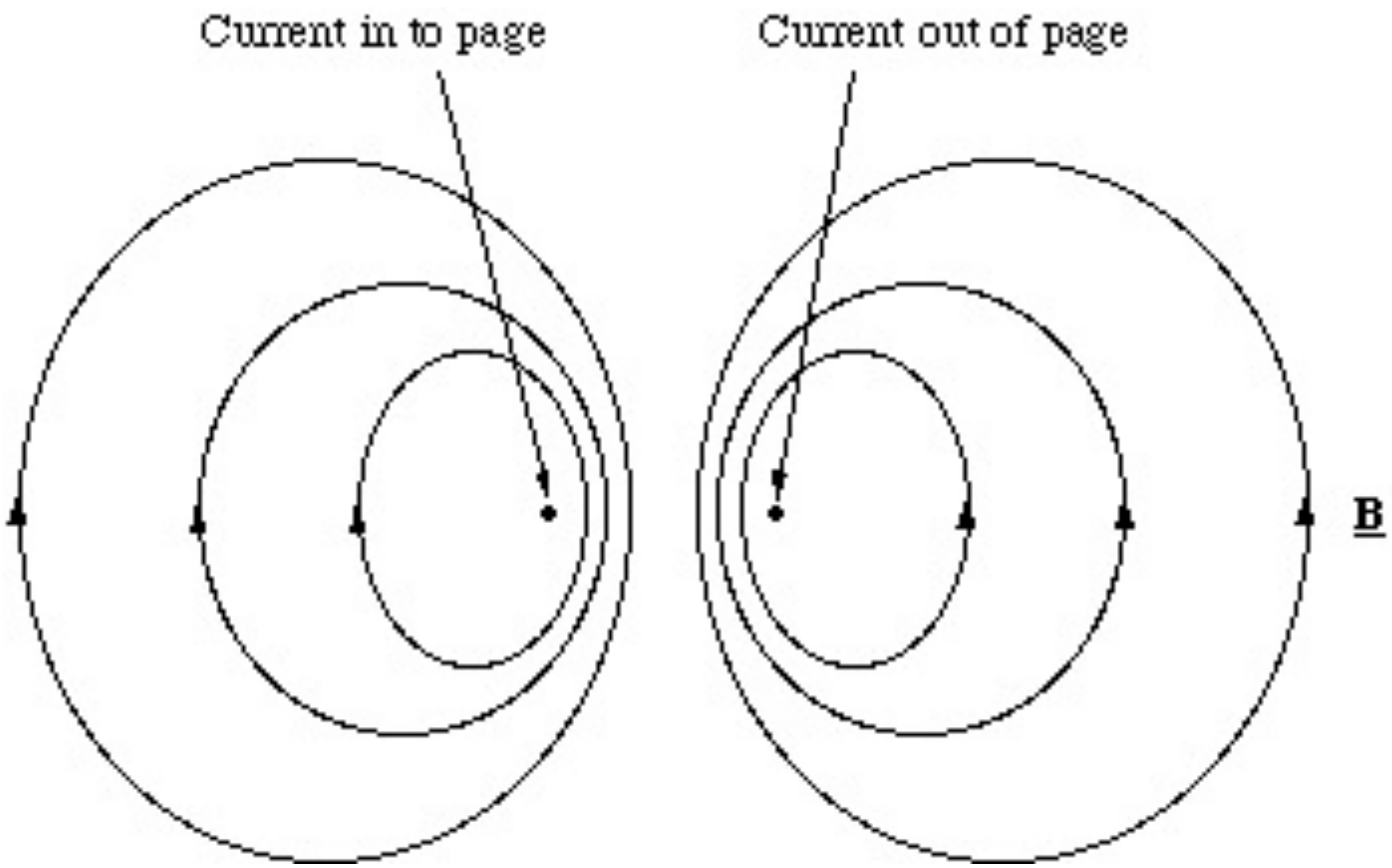
“Nuclear induction” and Nuclear Magnetic Resonance (NMR)

Magnetic dipole

1 Wire



A magnetic dipole can be obtained with 2 wires with currents flowing is opposite direction



Magnetic dipole

A magnetic dipole can also be obtained with a rotating charge

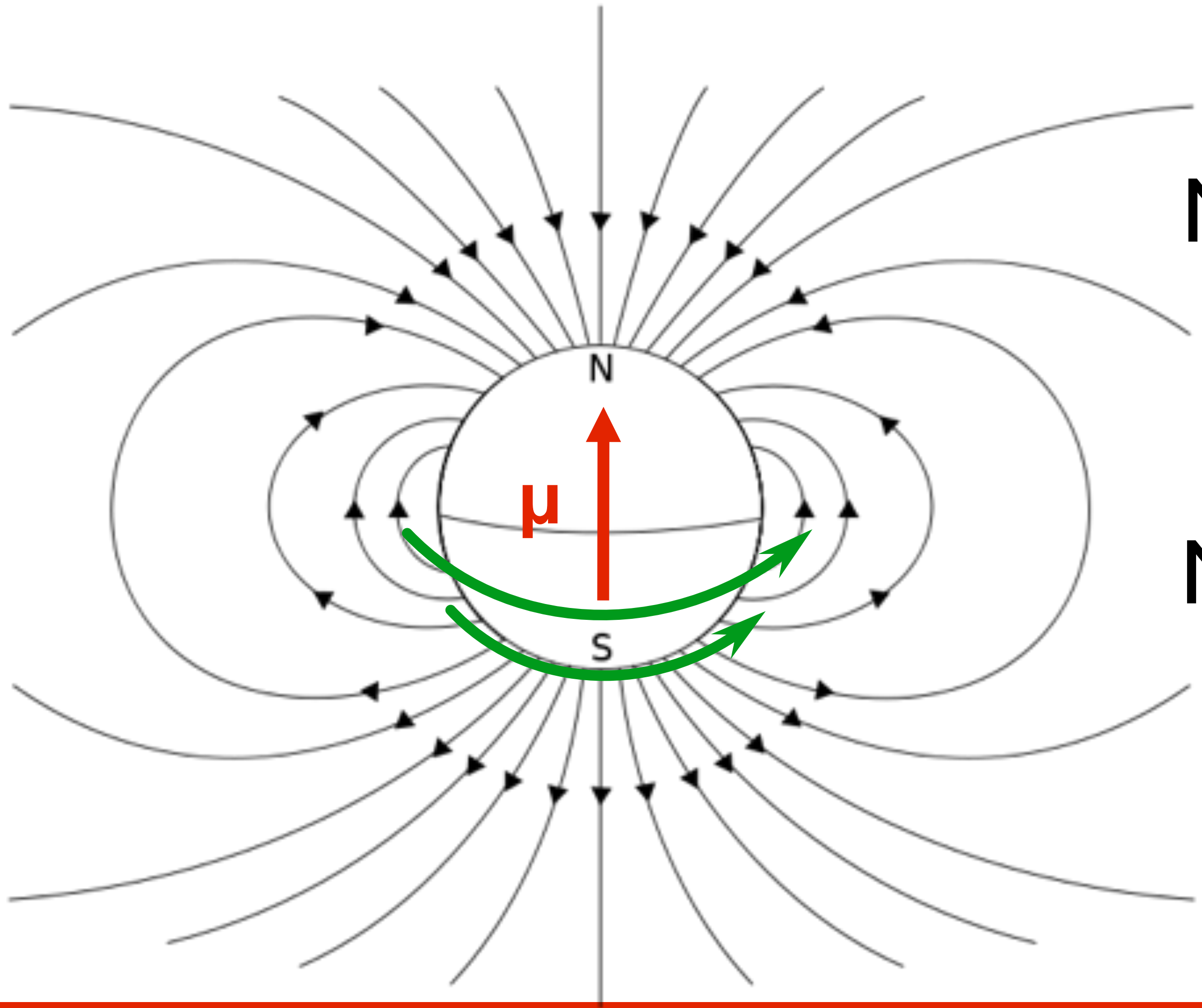
Rotation: $\vec{\Omega}$

Speed: $v = r\vec{\Omega}$

ρ_m Mass density

ρ_e Charge density

μ: magnetic moment (A/m)



Kinetic momentum:

$$\vec{P} = \int_V \rho_m r v dV = \int_V \rho_m r^2 dV \vec{\Omega}$$

Magnetic moment induced by dq:

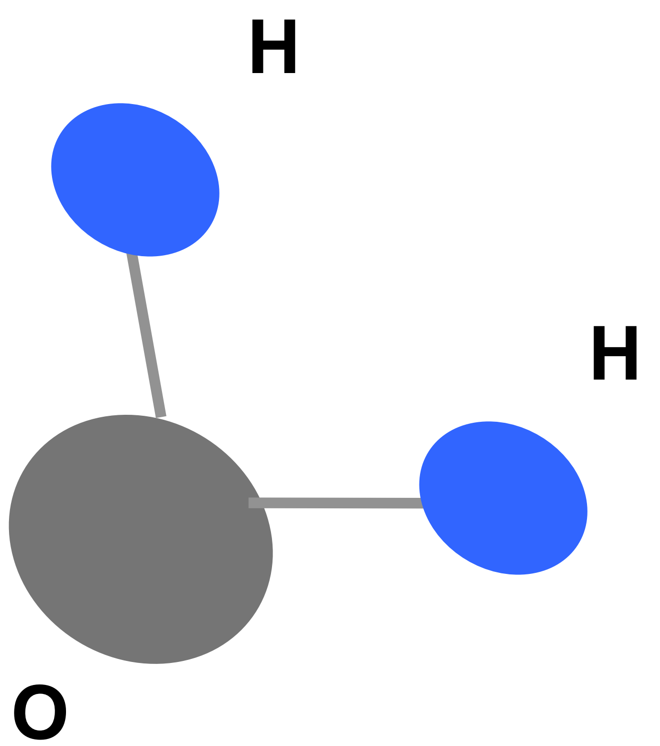
$$d\mu = \frac{1}{2} r v dq$$

Magnetic moment:

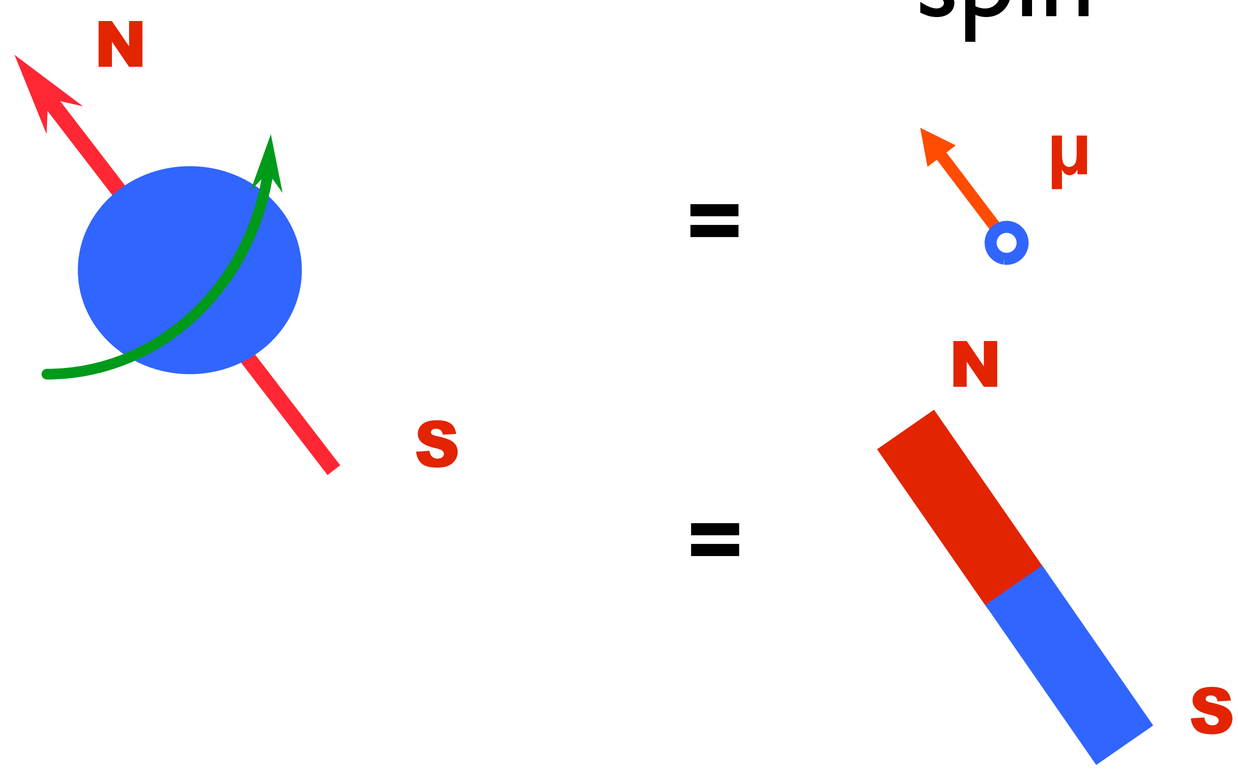
$$\vec{\mu} = \int_V \frac{1}{2} r v \rho_e dV = \frac{1}{2} \int_V \rho_e r^2 dV \vec{\Omega}$$

Hydrogen (H) nuclei

Water molecule



spin



"Like" a small magnet with a magnetic moment

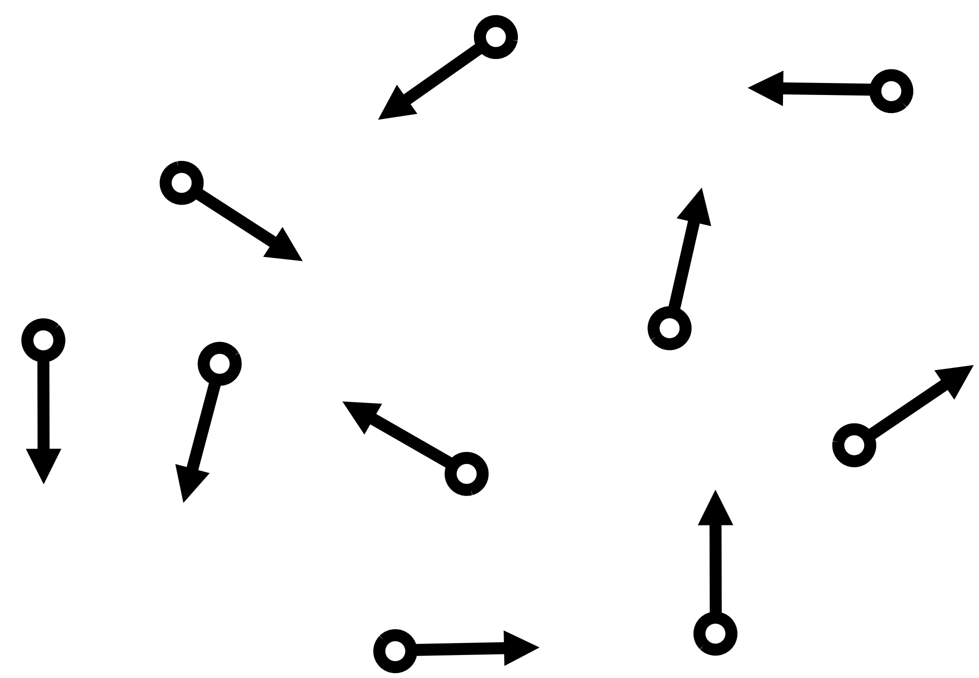
While a charge can rotate at different speed the **speed is fixed** and intrinsic for a spin. **Only the orientation can vary.**

Remark : NMR possible with other nuclei. You just need a spin i.e. impair number of protons + neutrons

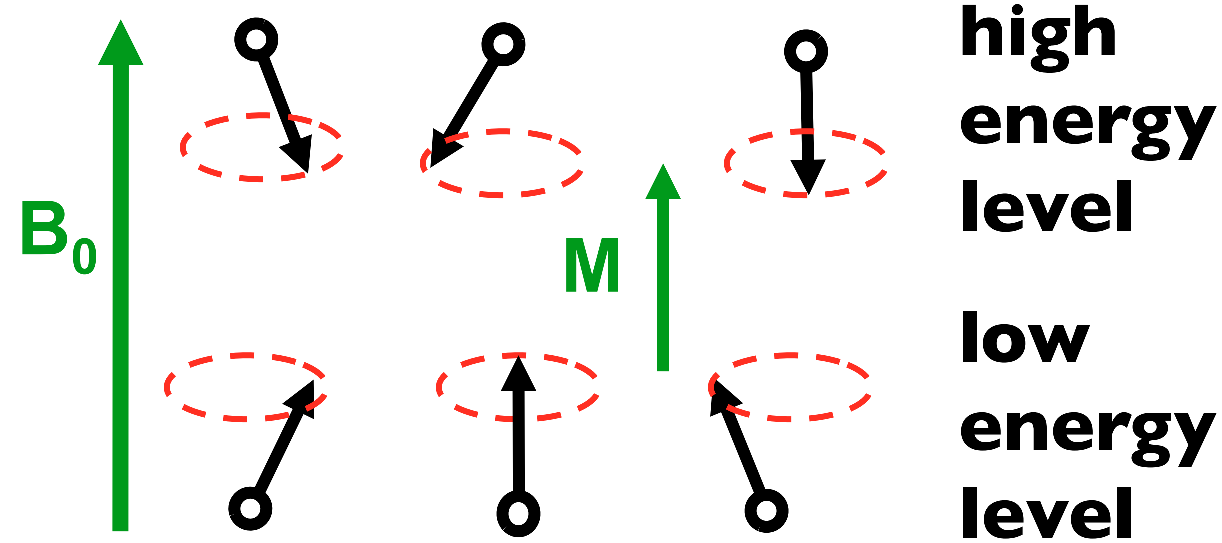
Spins in an external field

No field

spins with random orientation



In B0 field



$M = \sum \mu$ (linear)

Spins align with external field producing a global **magnetization M**

M is the magnetization **induced** by an external magnetic field

M depends on the **proton density**, the temperature, amplitude of B_0

Remark: Not all spins in low energy state: quantic effect, thermodynamic

A material can be...

paramagnetic: M aligns with B

diamagnetic: M tends to cancel the effect of B

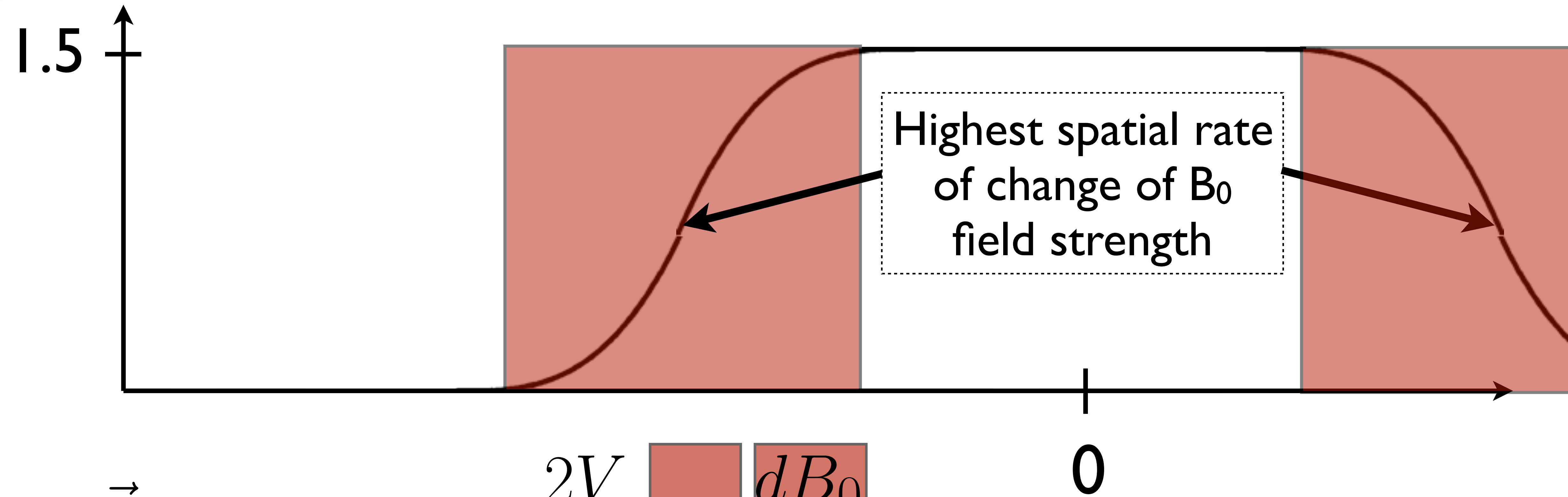
ferromagnetic: M is present even when B is absent (permanent magnet)

*What changes is the **magnetic susceptibility** of the materials*

Keep in mind: the magnetic susceptibility has an effect on the local magnetic field (what makes fMRI possible)

Ferromagnetic material in a scanner

Ferromagnetic material in spatial varying field



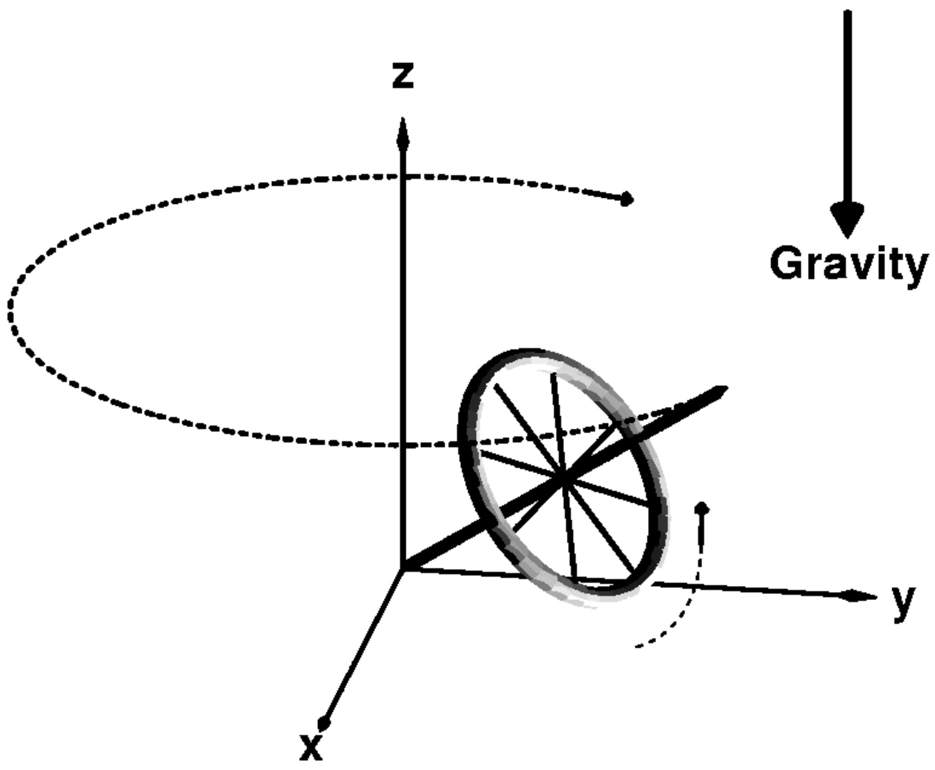
$$\vec{F}_{\text{mag}} = C_{\text{material}} \frac{2V}{\mu_0} B_0 \frac{dB_0}{dz}$$

This factor depends on shape & material:
~1 for a ferrous sphere
~500 for a ferrous cylinder!

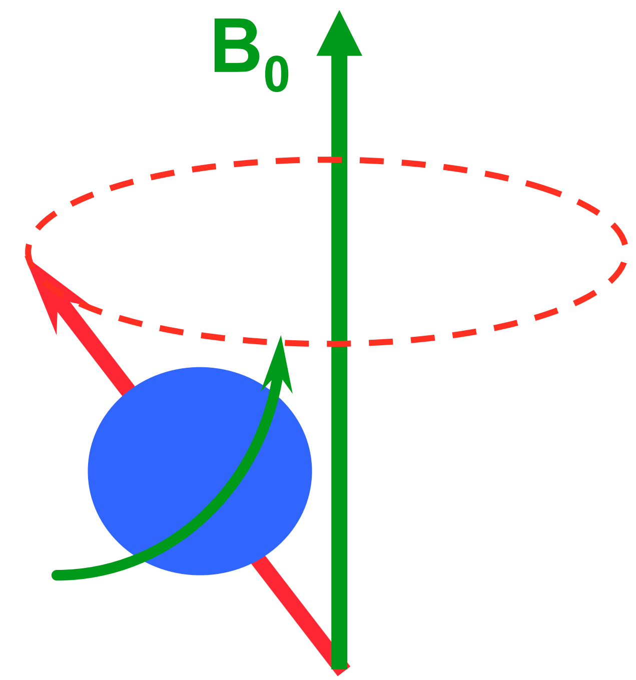
Depends on both field strength and spatial derivative!

Precession

Spinning top in a gravitational field



Magnetic moment in a magnetic field



$f_0 = \gamma B_0 / 2\pi$

$\omega_0 = \gamma B_0$

f_0 : Larmor frequency (63.86 MHz at 1.5 T for H)

A magnetic field induces a rotation around the direction of the field

Precession

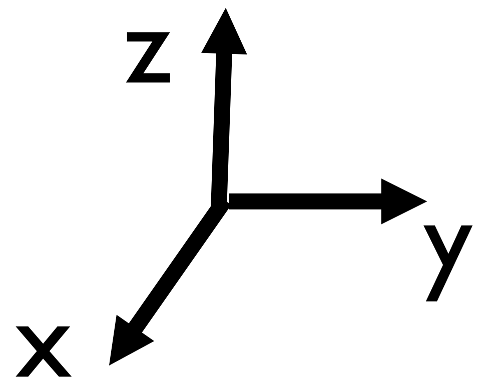
$$\omega_0 = \gamma B_0$$

- γ is the gyromagnetic ratio
- ω_0 depends on the field strength and the type of nuclei

Nuclei	γ (MHz / T)
^{1}H	42,575
^{13}C	10,7
^{31}P	17,235

A magnetic field induces a rotation around the direction of the field

Precession

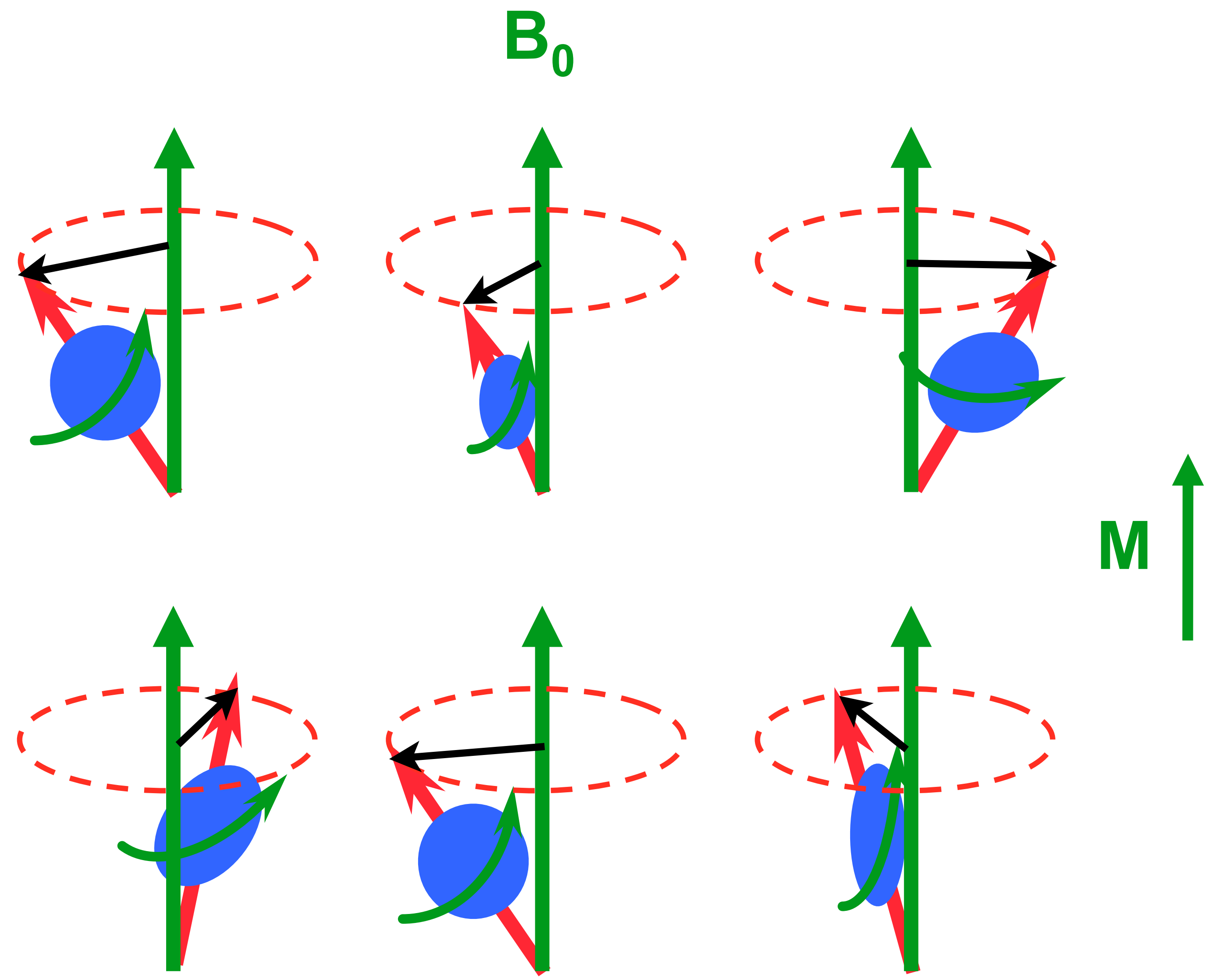


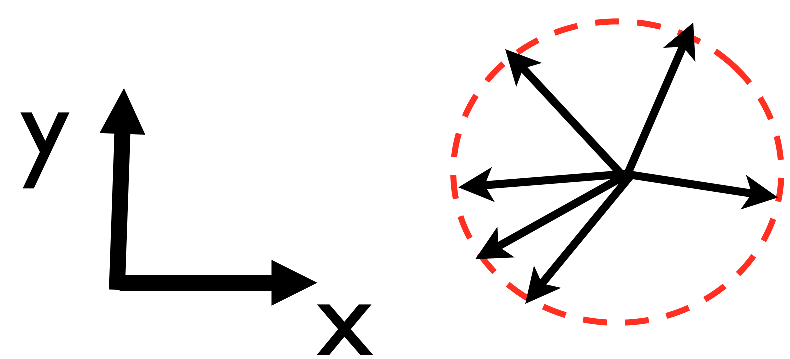
$M = M_z + M_{xy}$

M_z = longitudinal magnetization

M_{xy} = transverse magnetization

With B_0 only $M = M_z$
($M_{xy} = 0$)





Random phase

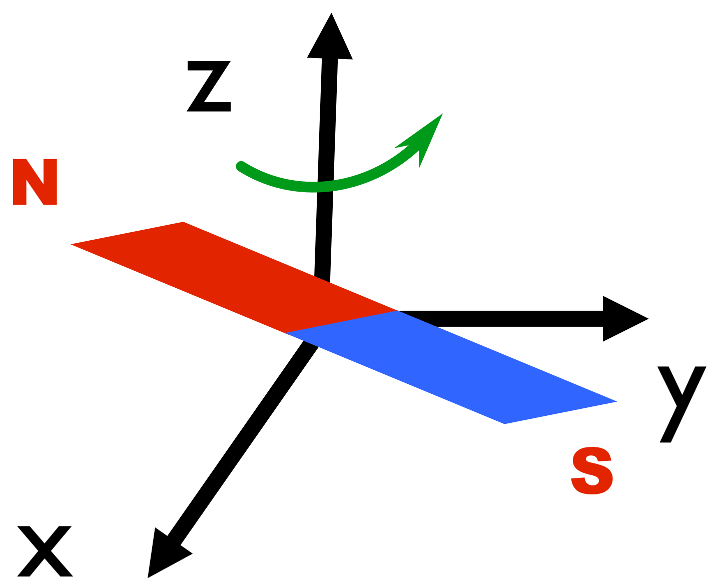
to sum up

- Spins precess in the direction of B
- Frequency depends on:
  - nuclei
  - field strength
- Spins are either down or up but more up than down to create the constant magnetization M
- Phases are random in the transverse plane and produce no M_{xy} magnetization

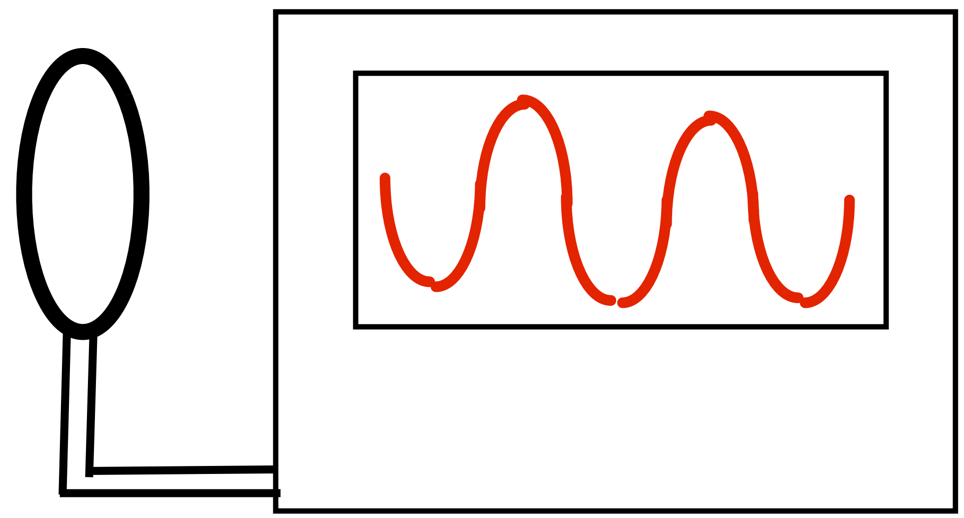
The idea of NMR is to perturb the constant magnetization to produce a signal

Induction

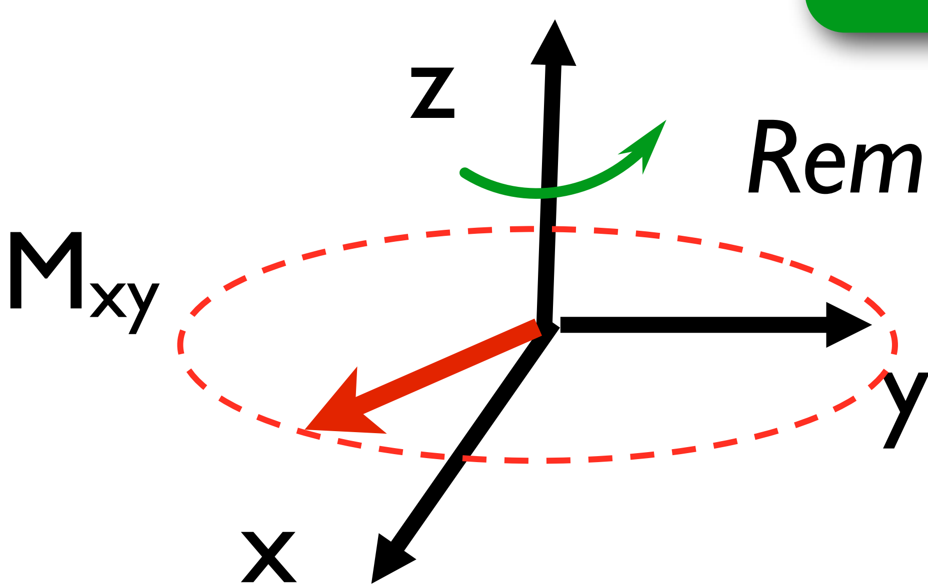
Rotate
a magnet



Induced current



Changing magnetic flux through a coil induces a detectable difference of potential



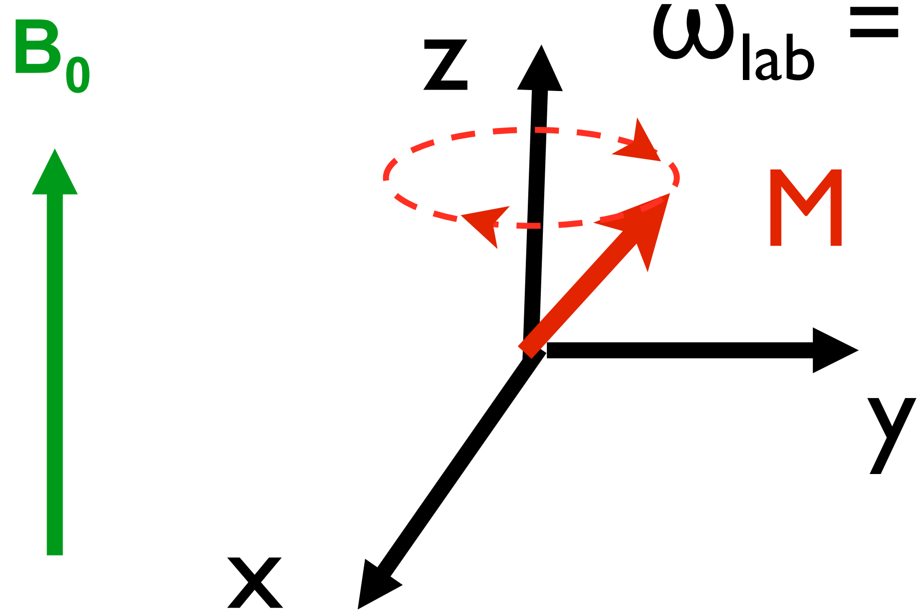
Remark: Same idea as the dynamo on a bike

NMR uses M_{xy} as a rotating magnet : it's "Nuclear induction"

Frames of reference

Lab frame

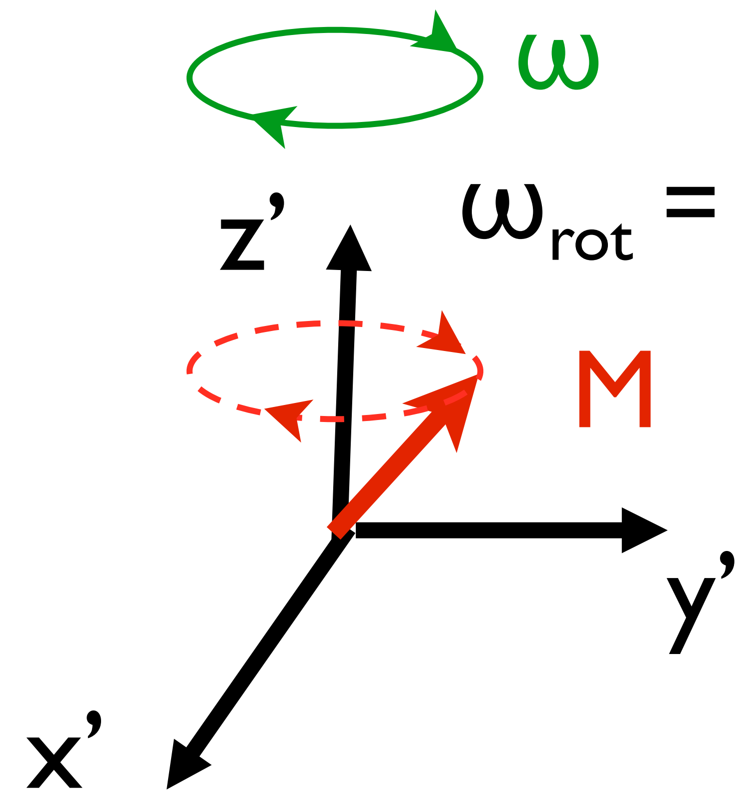
$$\omega_{lab} = \omega_0 = \gamma B_0$$



Magnetization precesses rapidly in lab frame

Rotating frame

$$\omega_{rot} = \omega_0 - \omega$$

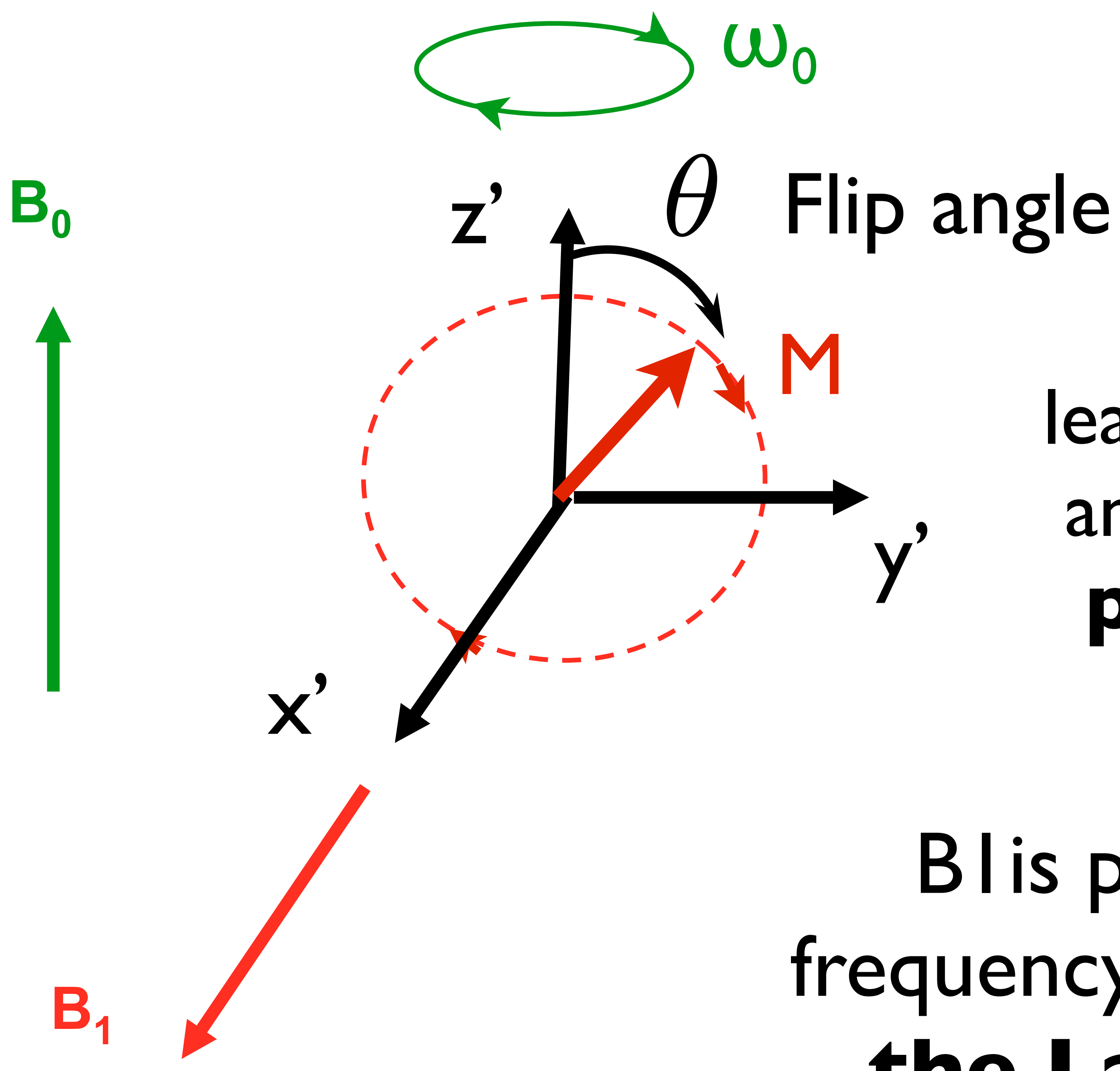


Magnetization precesses slowly in rotating frame

Special Case: If frame rotates at Larmor frequency, magnetization appears stationary!!!

B1 and radio-frequency (RF) pulse

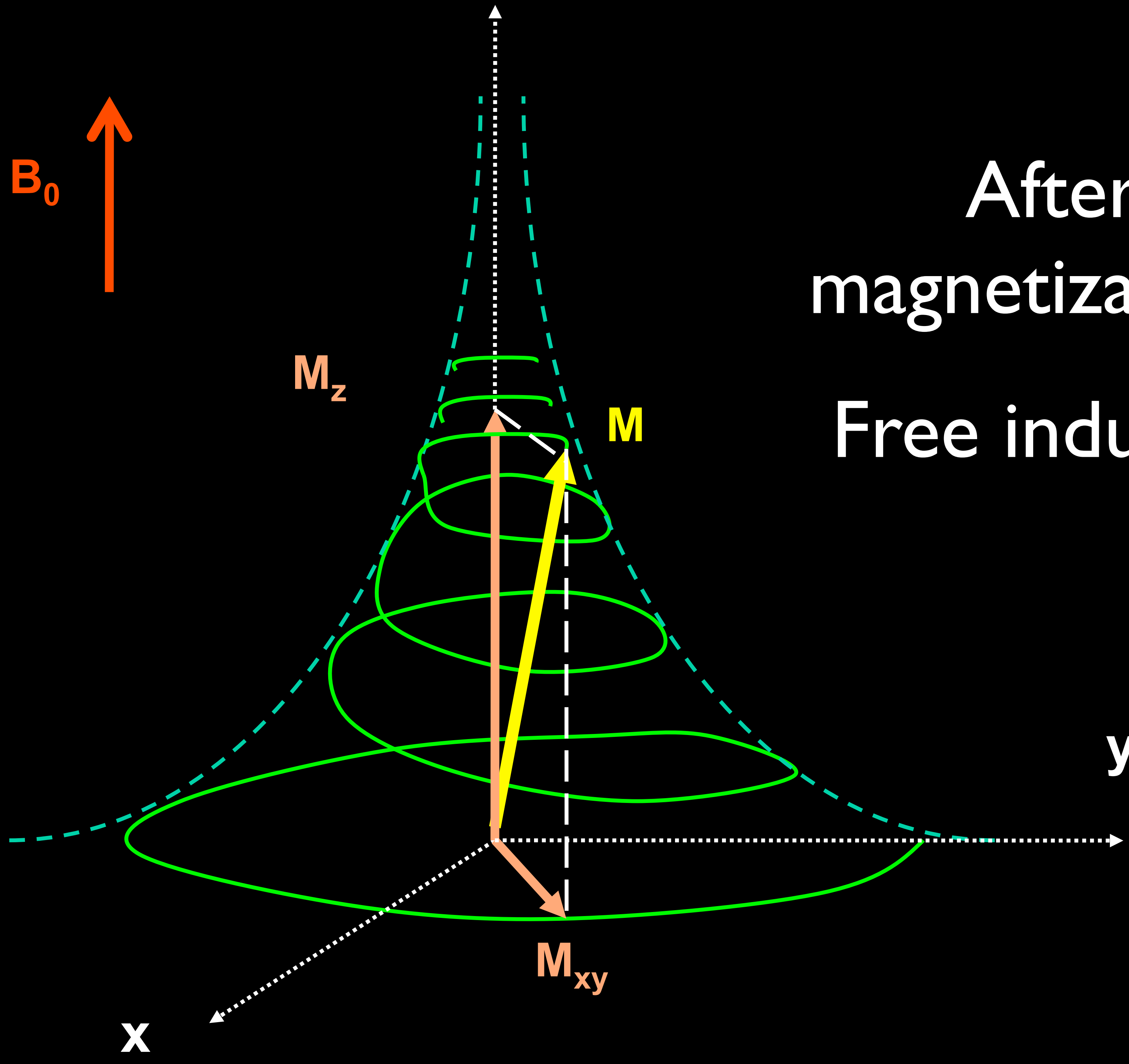
Rotating frame with stationary magnetization



B1 field along x'
leads to a precession around x'
and rotates the magnetization
producing a transverse magnetization

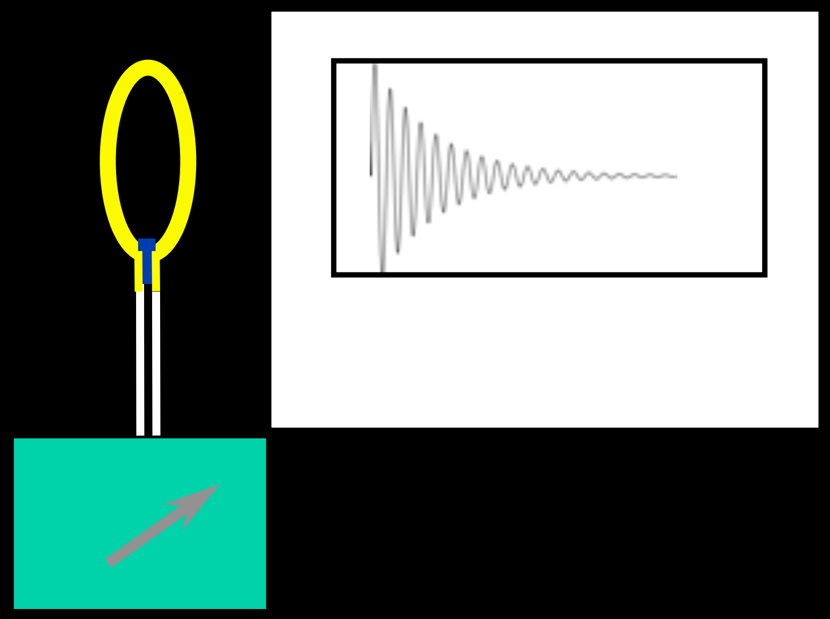
B1 is produced by a radio frequency (RF) pulse **tuned at the Larmor frequency**

Longitudinal Relaxation: T1



After a 90° flip the magnetization returns to M_0

Free induction decay (FID)

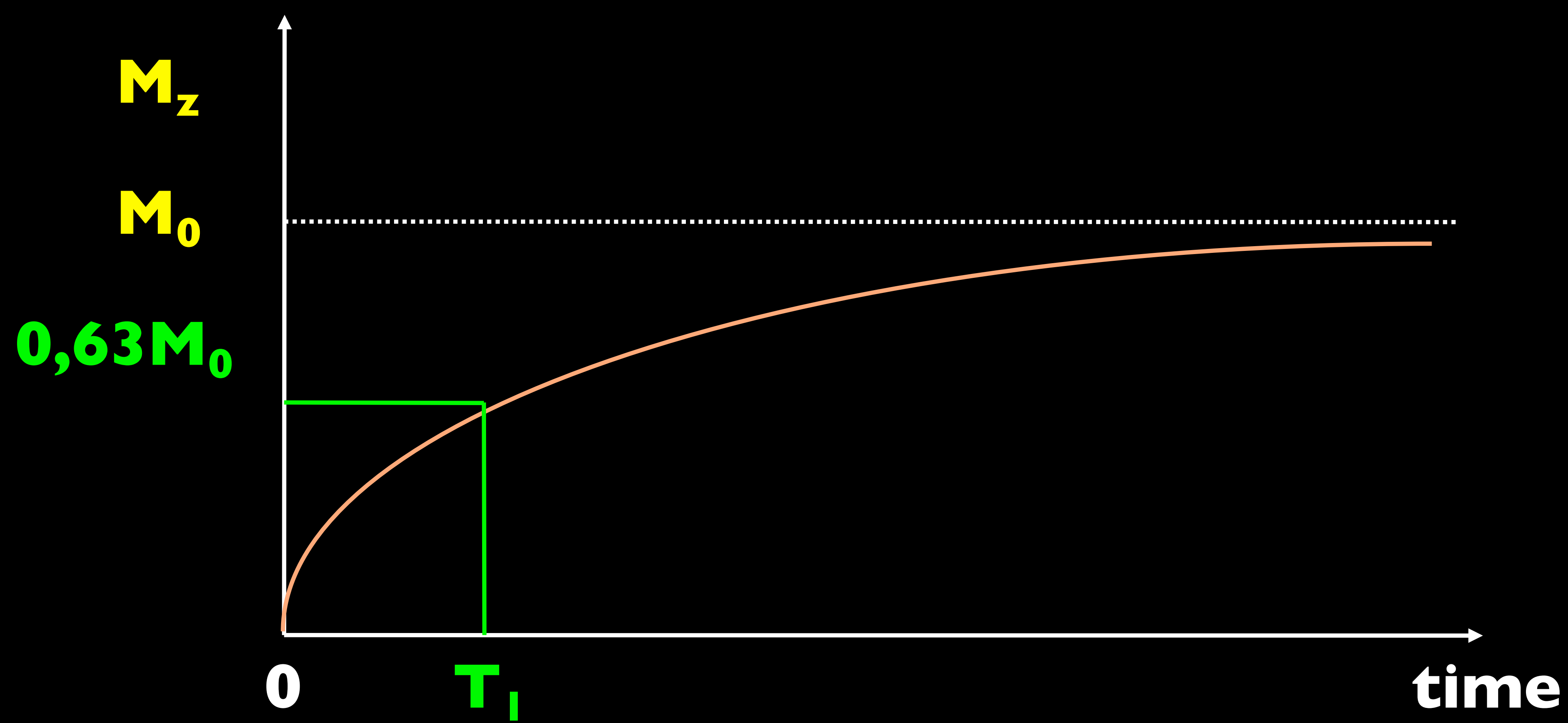


Longitudinal Relaxation: T1

1st Bloch's law after 90° flip: $M_z(t) = M_z(0)(1 - e^{\frac{-t}{T_1}})$

T1 : longitudinal relaxation time

Due to B_0 field

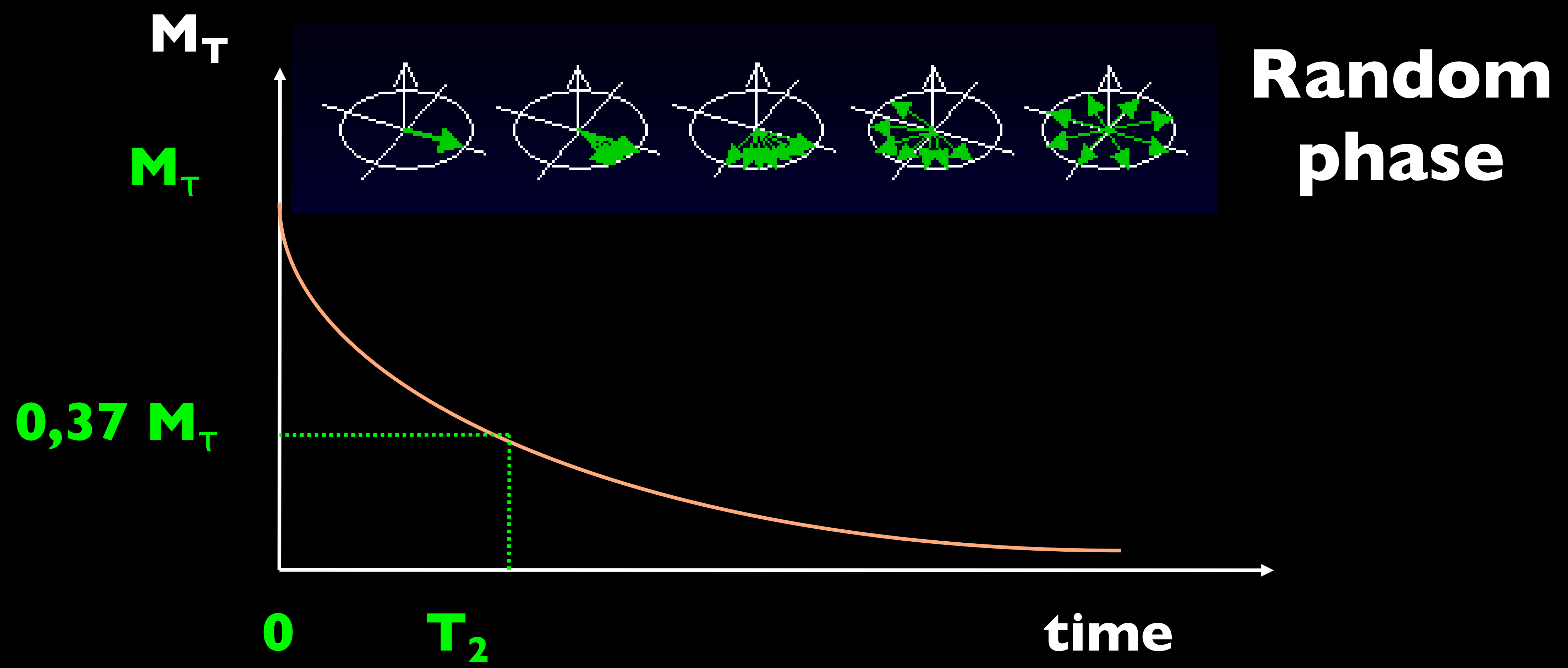


Transverse Relaxation: T2

2st Bloch's law after 90° flip: $M_{xy}(t) = M_{xy}(0)e^{-\frac{t}{T_2}}$

T2 : transverse relaxation time

Due to progressive dephasing of the spins



Relaxation

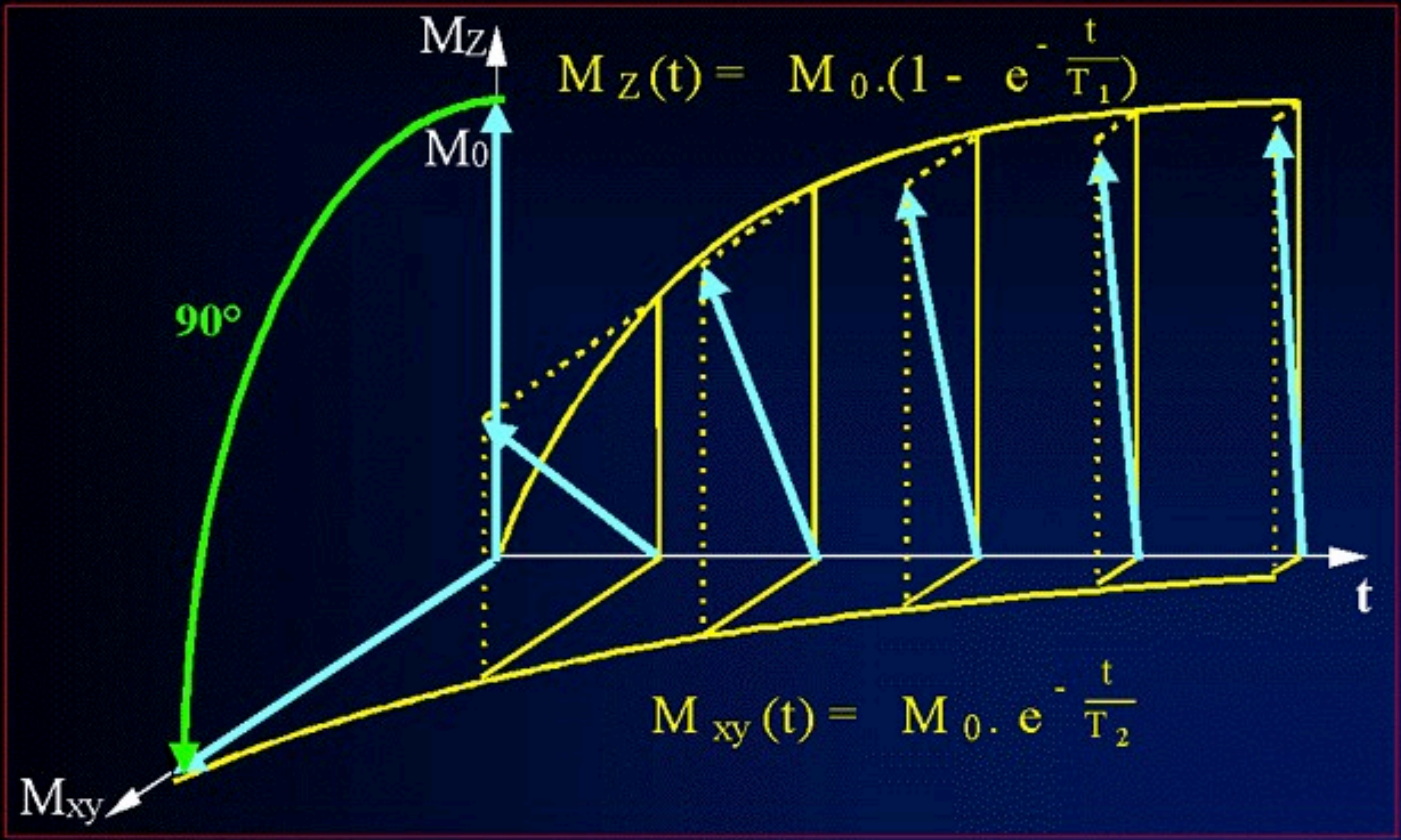
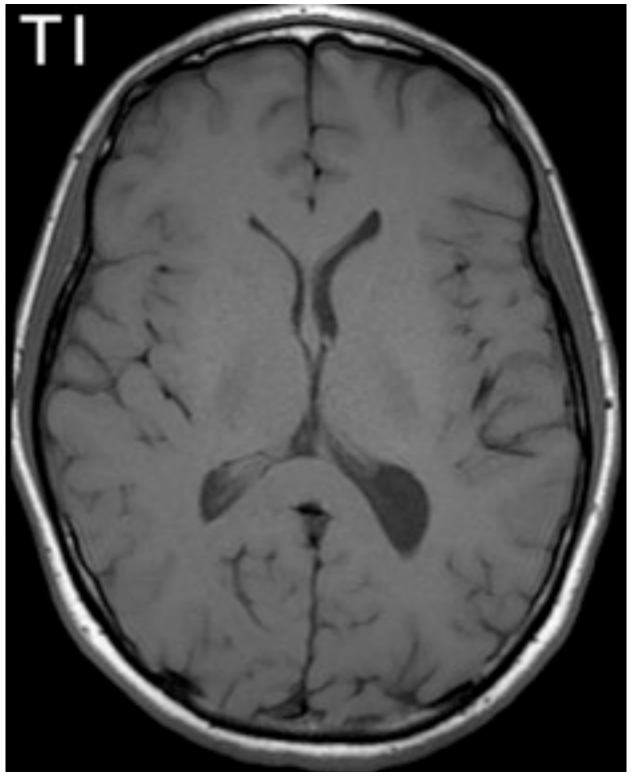
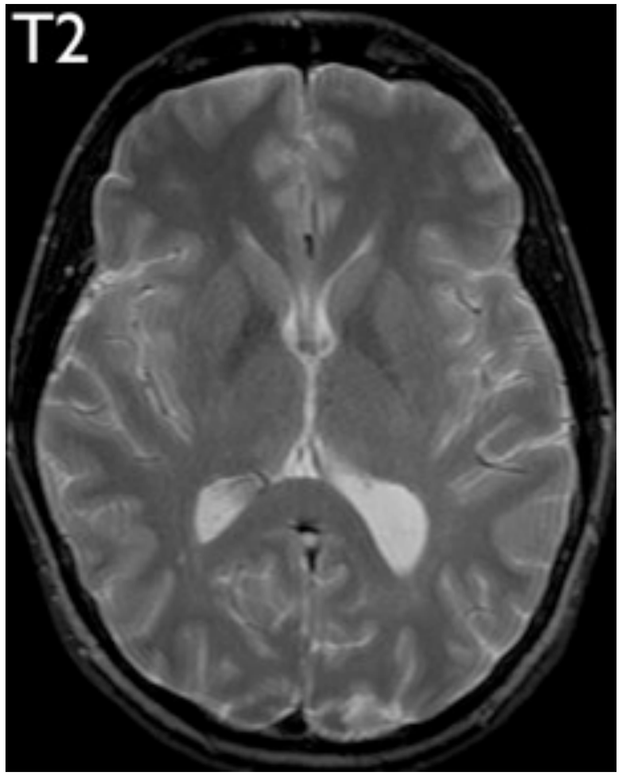


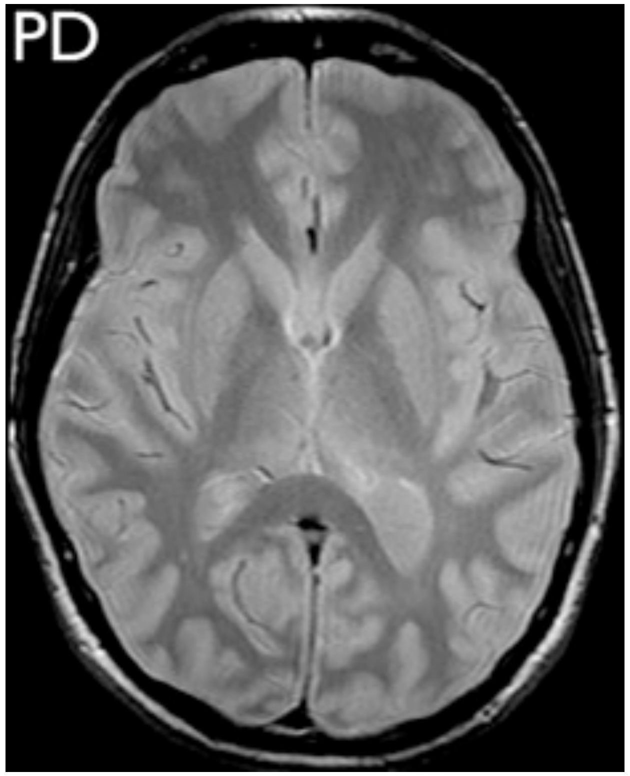
Image contrasts



T_1-weighted



T_2-weighted



PD-weighted

- Image contrast = signal difference between different tissues of interest
- Image contrast is the combined effects of tissue differences in proton density (PD), T1 and T2
  - Many sequences can be either PD or T1 or T2 weighted, depending on details of timing
  - Typically aim to maximize the contrast due to one of these and minimize the contrast due to the others

T1, T2, proton density (PD)

Tissues have different magnetic properties:

	T_1 (1,5 T)	T_2
water	3000 ms	1500 ms
CSF	2500	1000
muscle	800	45
fat	200	75
white matter	750	90
gray matter	850	100

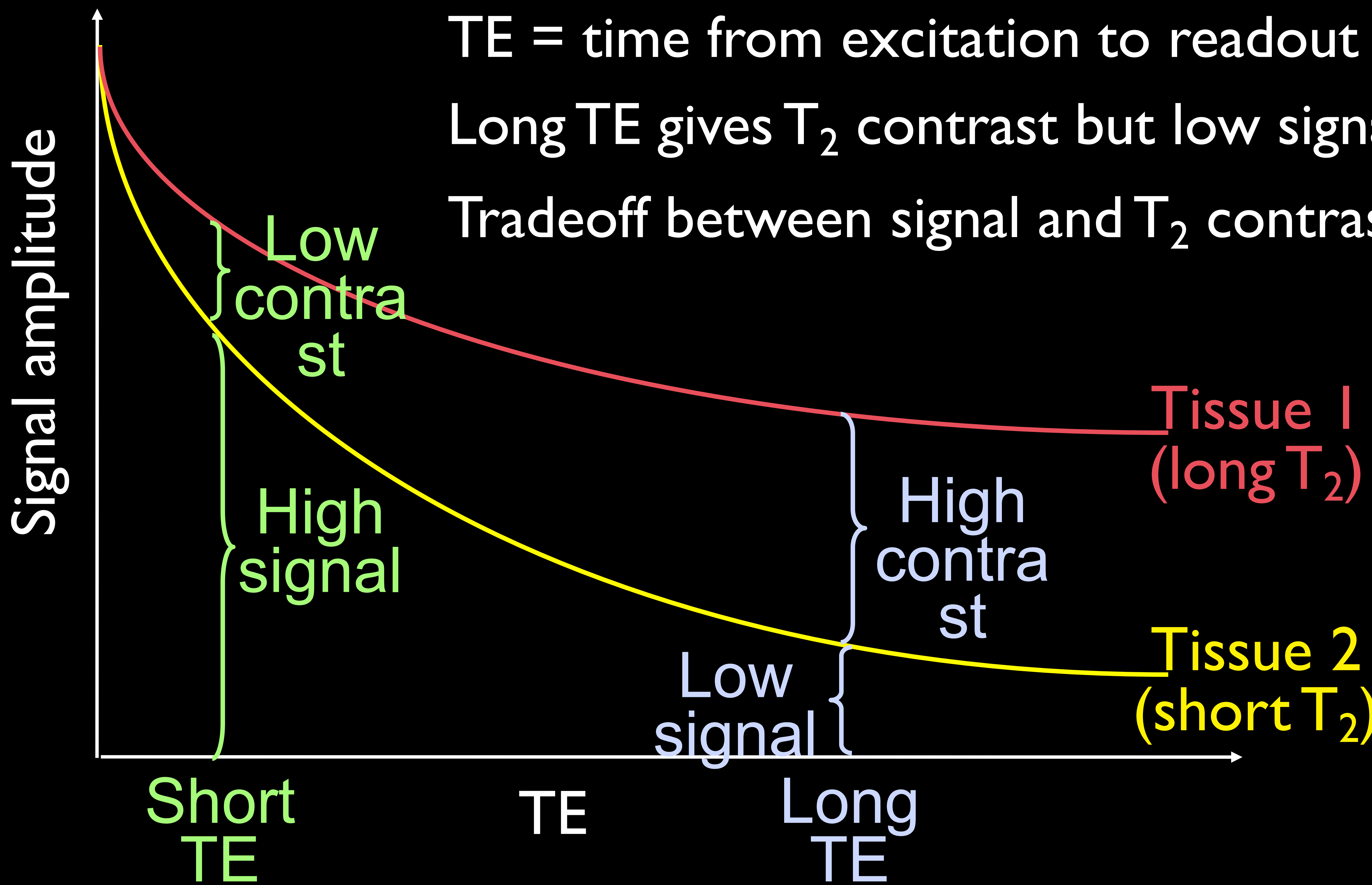
That's why you can see the difference

Echo time (TE) and T_2 contrast

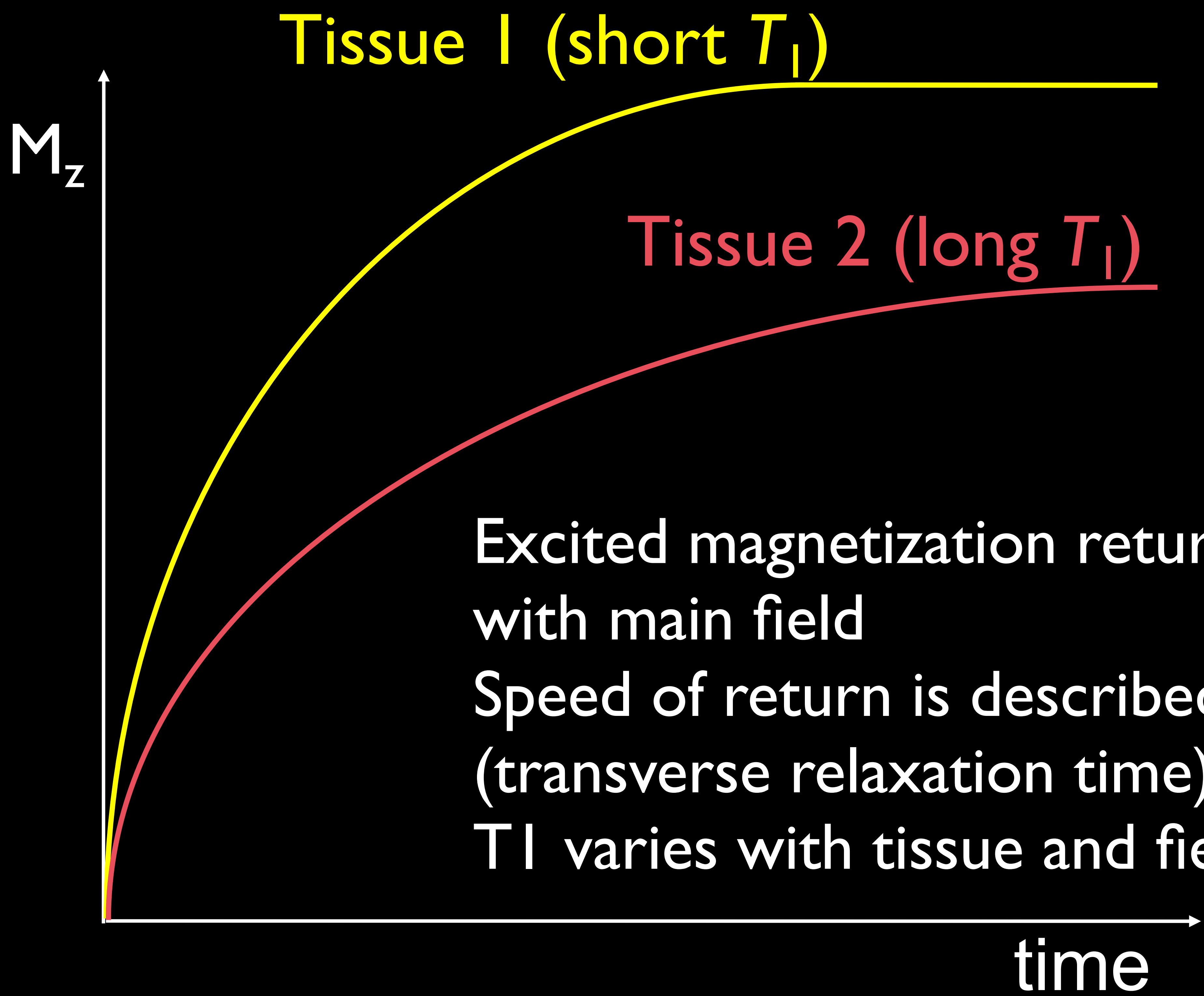
TE = time from excitation to readout

Long TE gives T_2 contrast but low signal

Tradeoff between signal and T_2 contrast



T_1 contrast

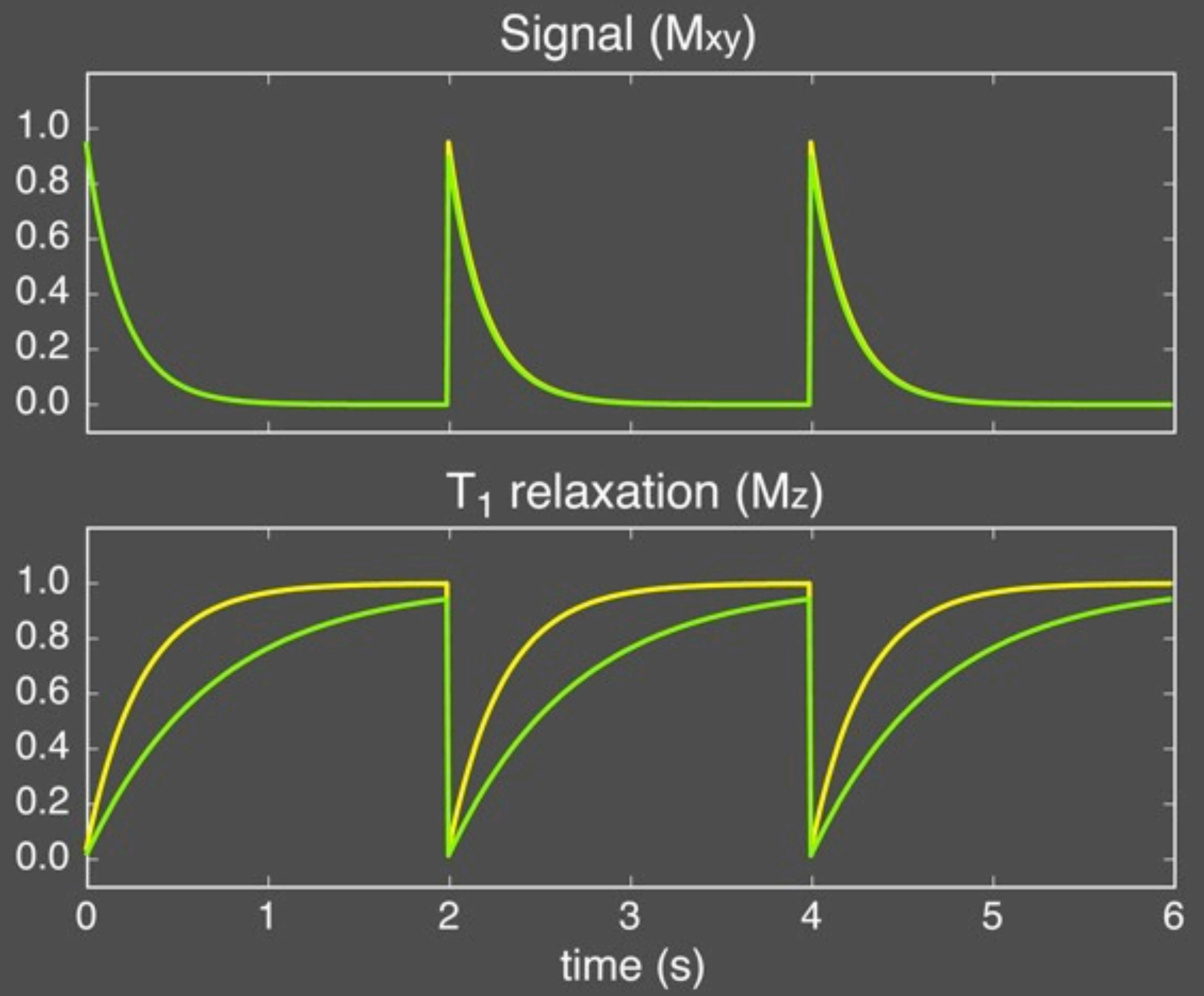


Excited magnetization returns to alignment with main field

Speed of return is described by T1 (transverse relaxation time)

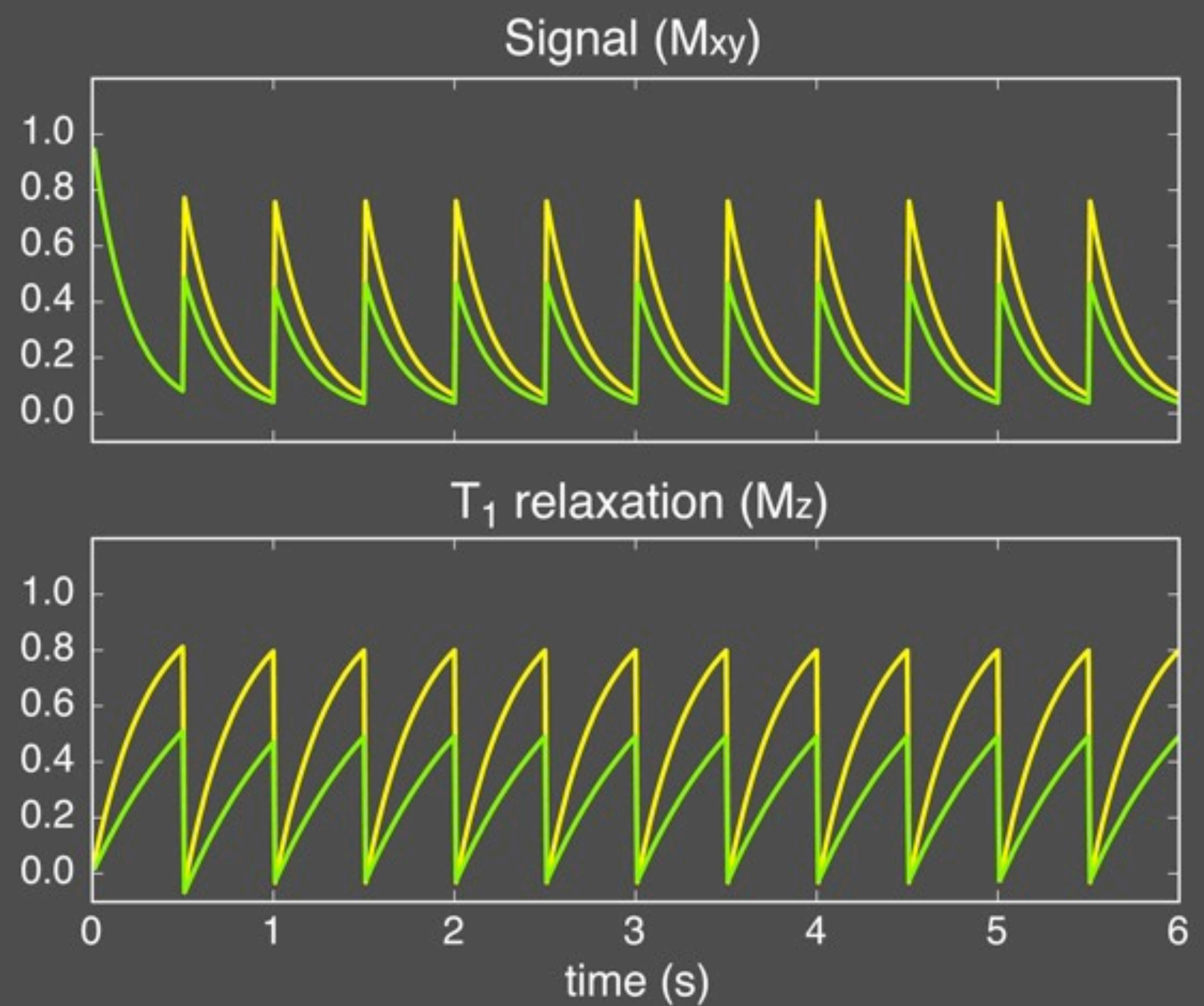
T1 varies with tissue and field strength

TR and T_1 Contrast



Long TR: T_1 recovers completely for both tissues

TR and T_1 Contrast



Short TR: short T_1 recovers more T_1 contrast!

What's My Contrast?

	Short TE	Long TE
Short TR	T_1-weighted	~~T_1 *and* T_2 weighted~~
Long TR	Proton density	T_2-weighted

T2*

In practice we don't observe T2 but T2* (T2* < T2)

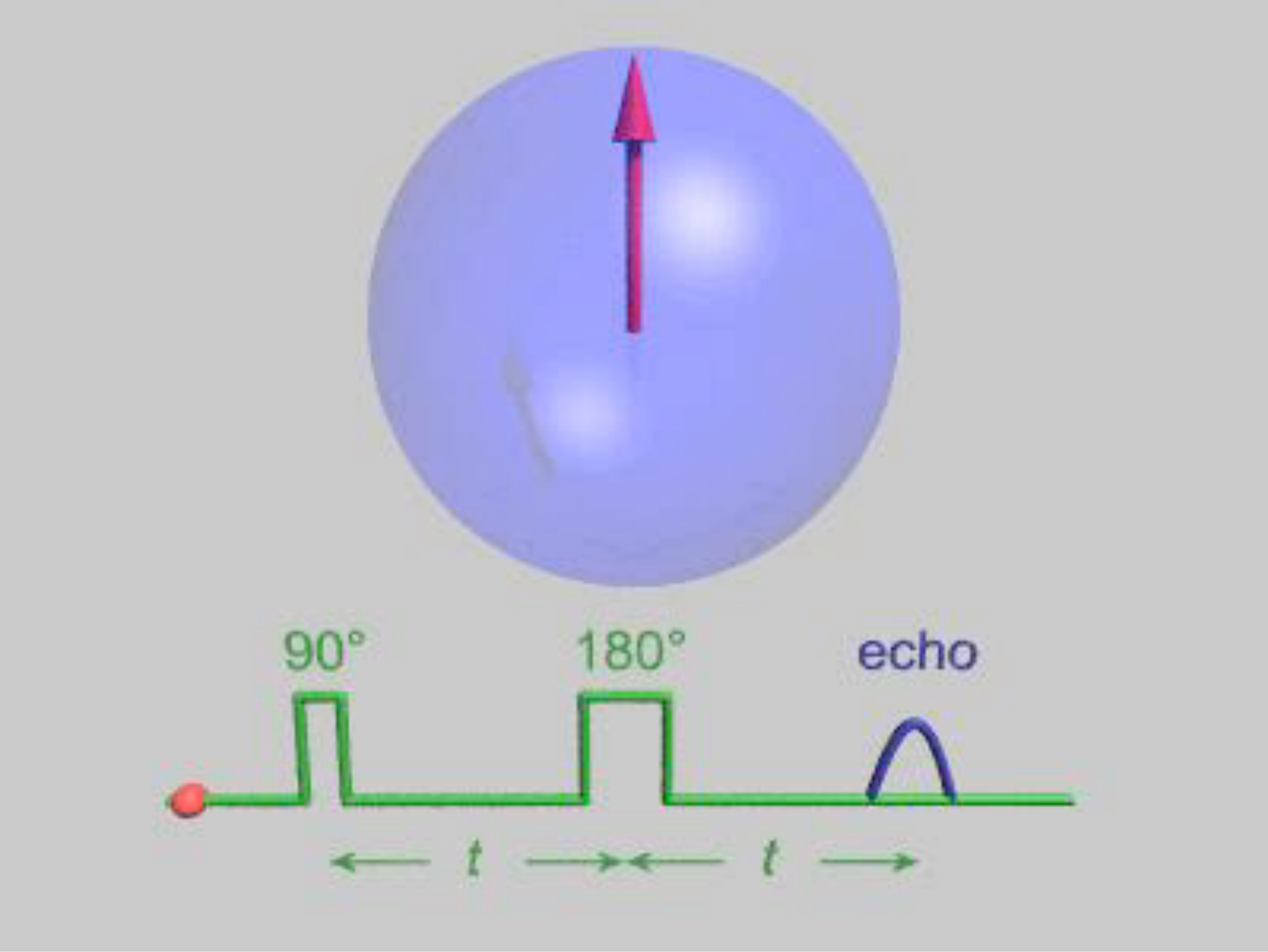
$$\frac{1}{T_2^*} = \frac{1}{T_2} + \frac{1}{T_2'} = \frac{1}{T_2} + \gamma \Delta B_0$$

It describes the exponential decay of signal, due to **spin-spin interactions**, **magnetic field inhomogeneities**, and **susceptibility effects**.

It's a problem for anatomical images but this "issue" is actually a blessing for functional and diffusion imaging.

Spin Echo

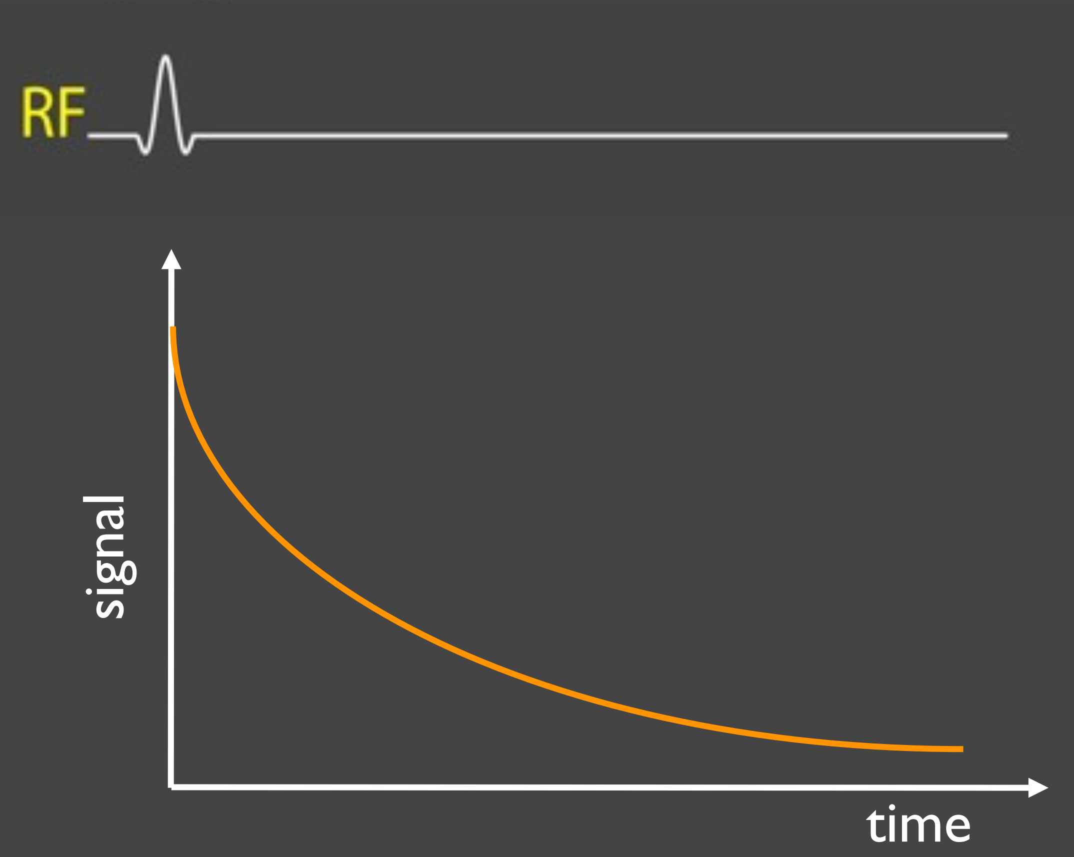
Objective : correct for T2* dephasing



[Erwin Hahn 1950]

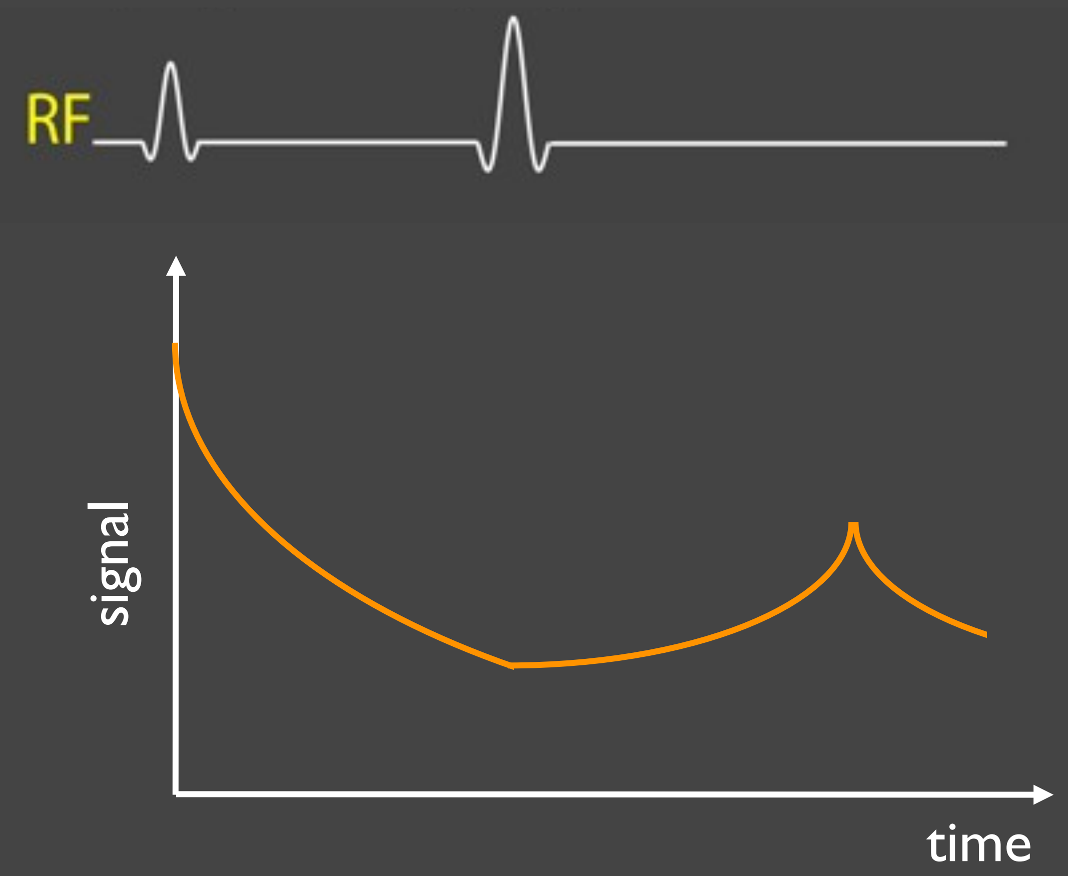
source: wikipedia.org

Spin Echo vs. Gradient Echo signal



GRE signal

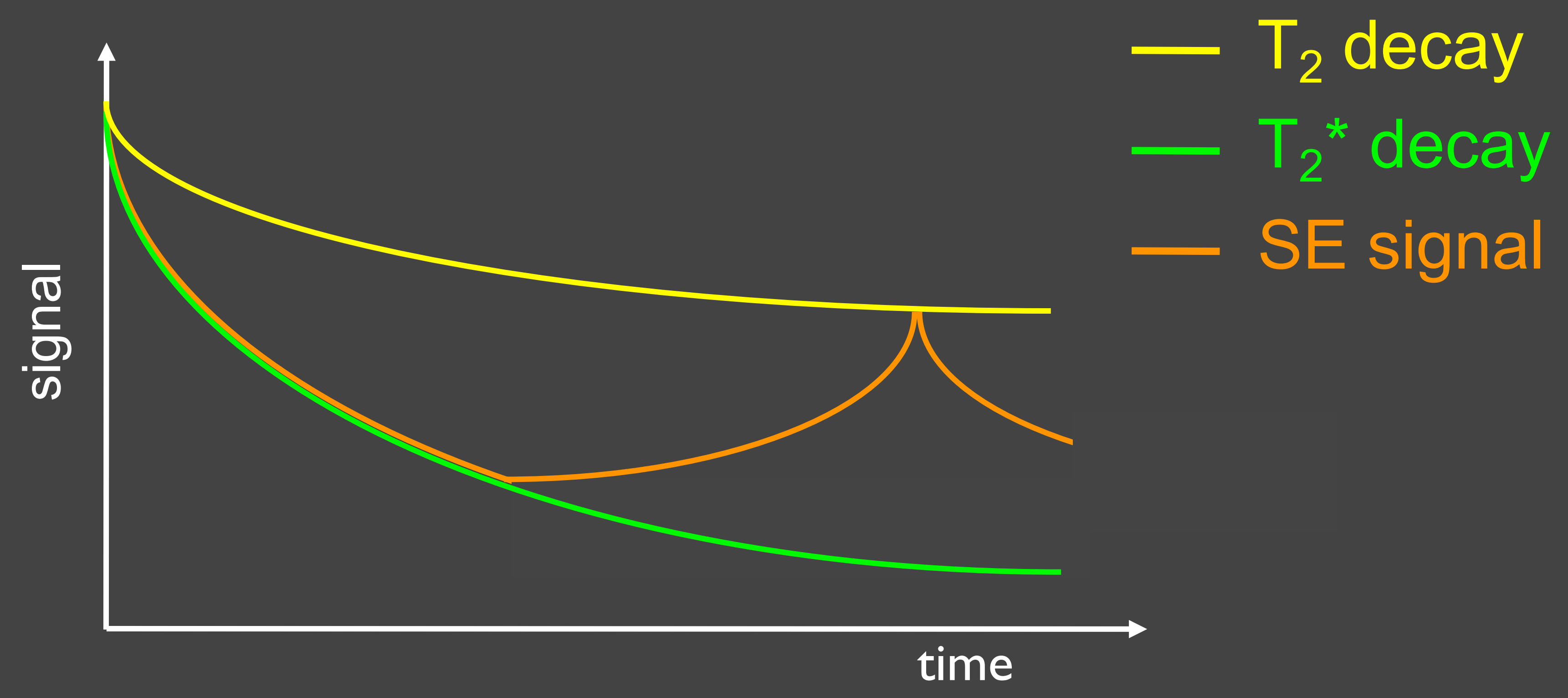
(more properly, a Free Induction Decay = FID)



SE signal

(signal decays, then comes back as "echo")

T_2 vs T_2* Relaxation



SE can refocus only part of the signal decay

- T_2* refers to part that can be refocused
- Without refocusing, signal will have T_2* contrast

Even spin echo signal experiences some decay

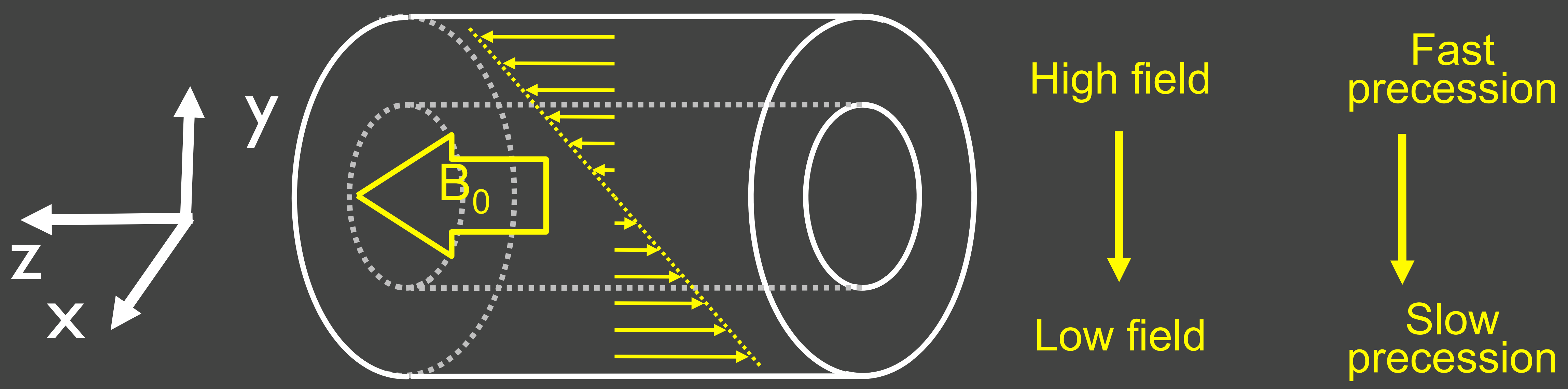
- T_2 refers to signal decay that cannot be refocused
- With refocusing, signal will have T_2 contrast

Magnetic Resonance **Imaging**

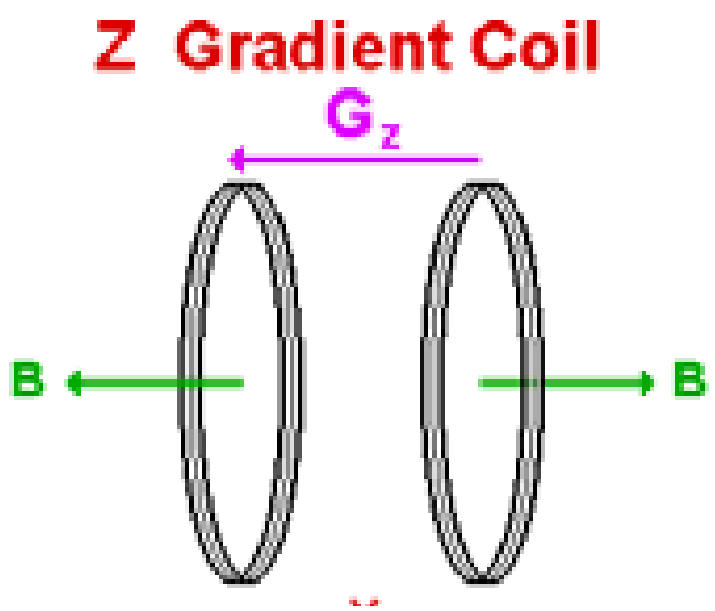
Magnetic gradient

Gradient: Extra magnetic field varying over space

– Precessional frequency varies with position!
– Detect signal at a position by selectively listening to the corresponding frequency
– "Pulse sequence" modulates size of gradient
– Gradient around 10 mT / m



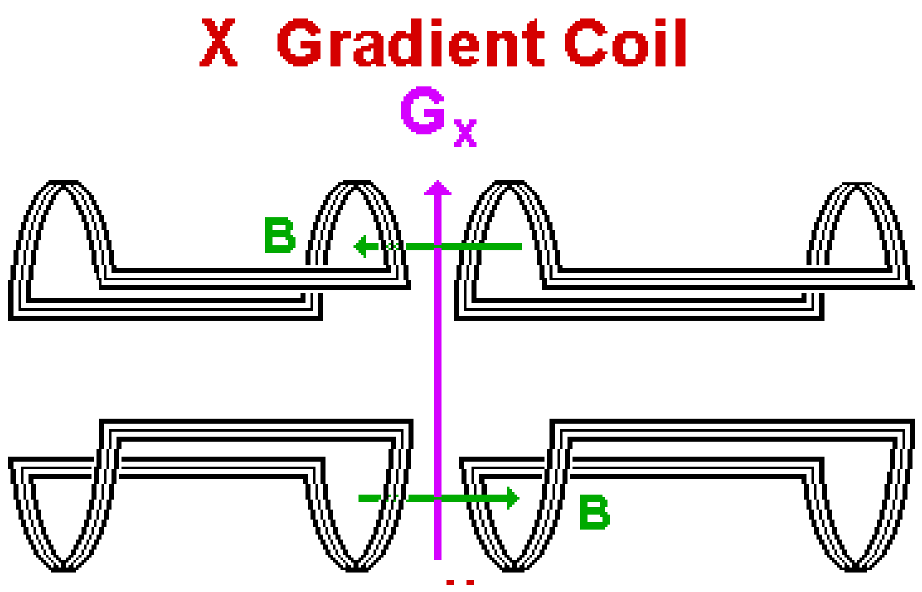
Magnetic gradients

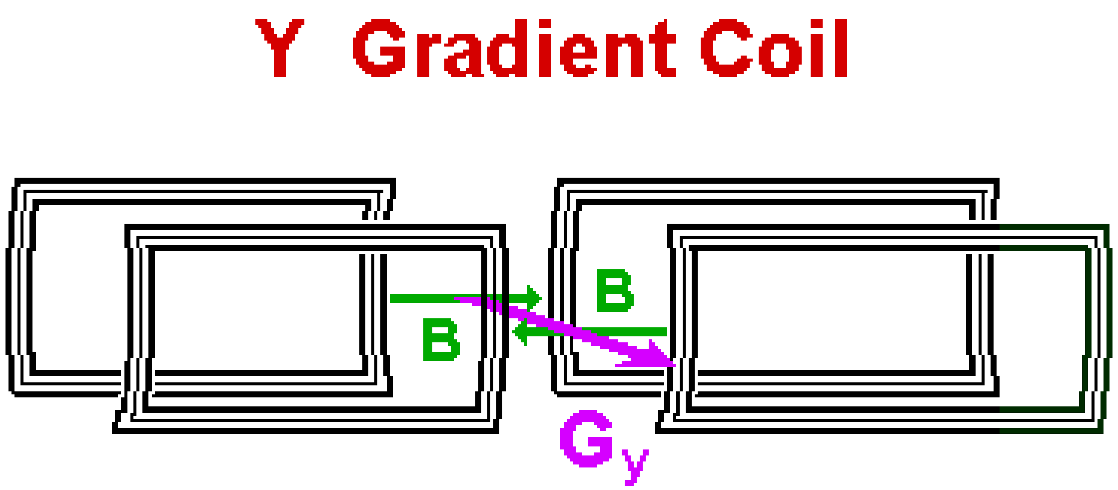


3 gradients in MR system

Fast gradients lead to fast imaging

Strong gradients lead to higher spatial resolution



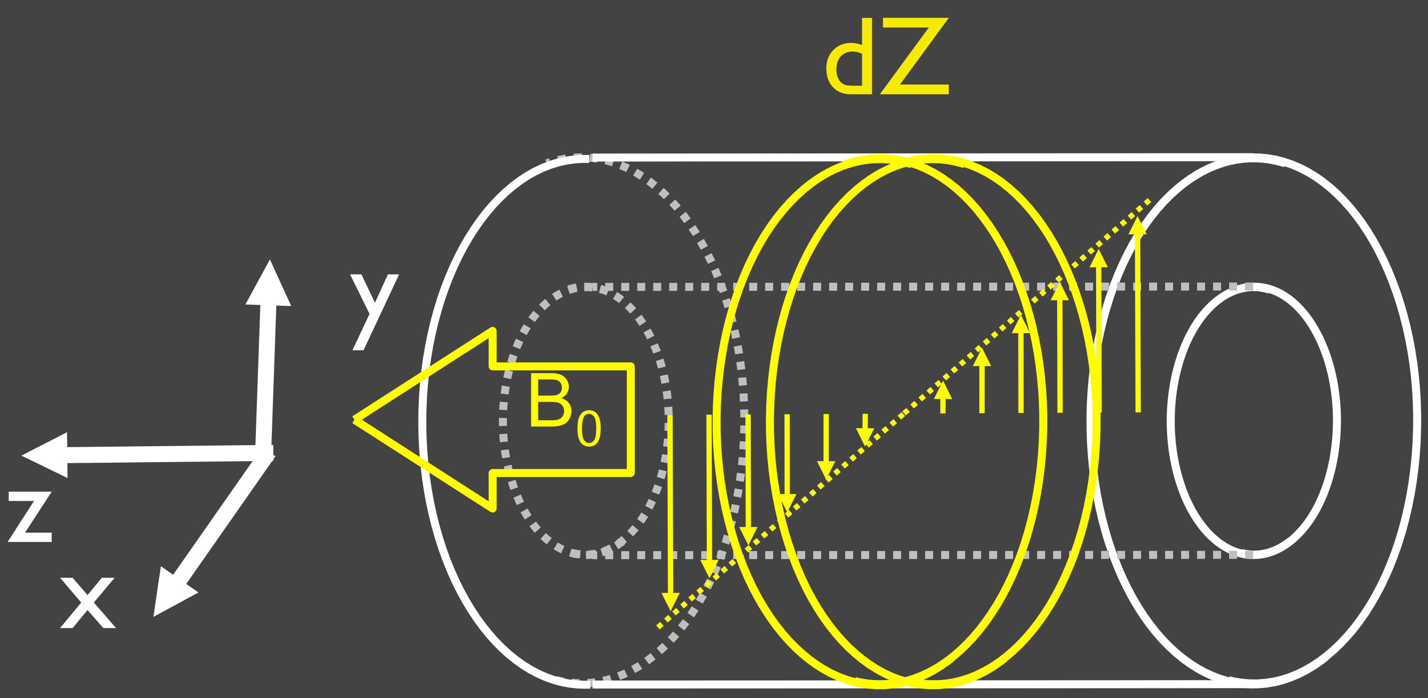


Slice selection

Gradient along z axis so field varies along z axis:

$B = B_0 + zG_z$

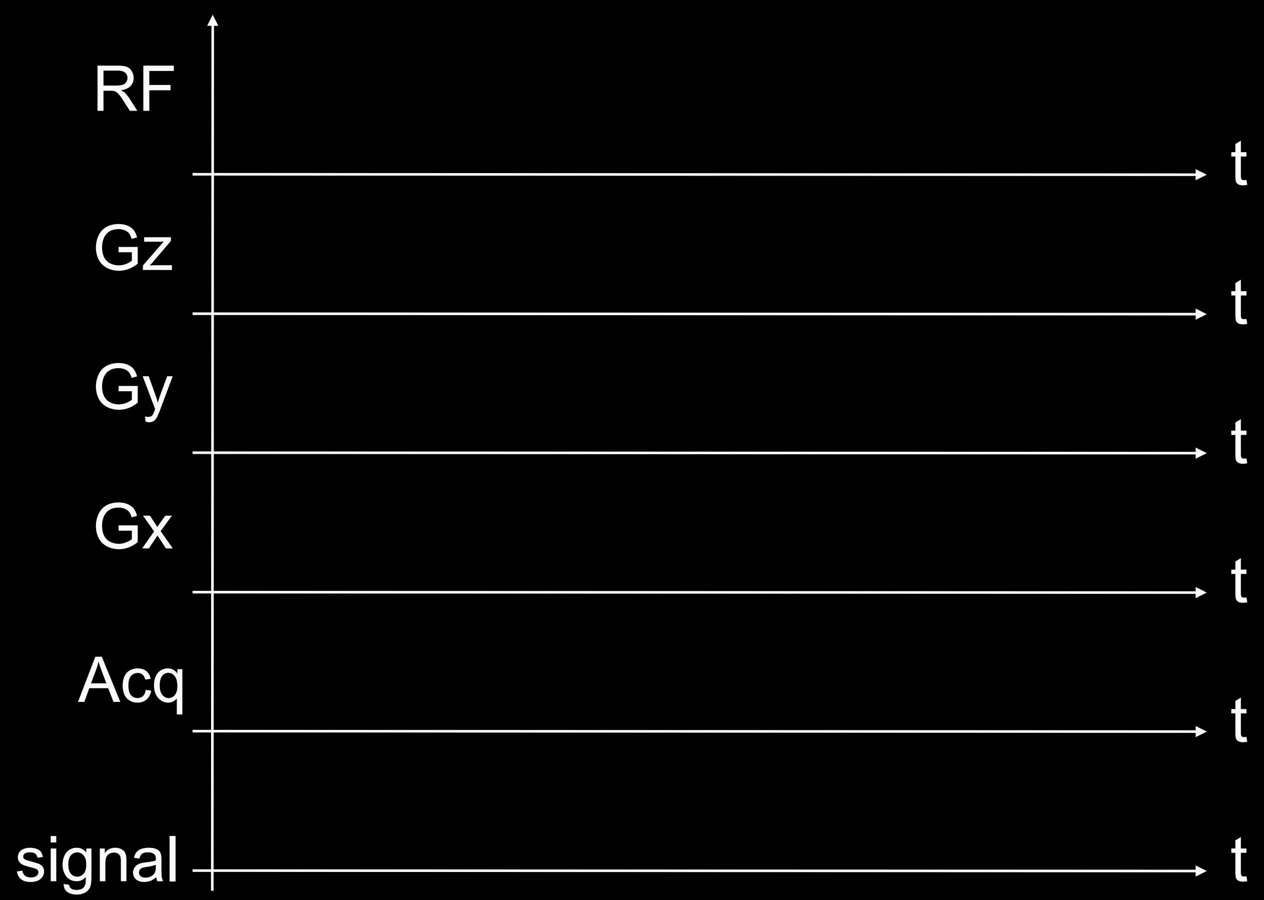
Field strength matches RF pulse frequency only for a slice dZ

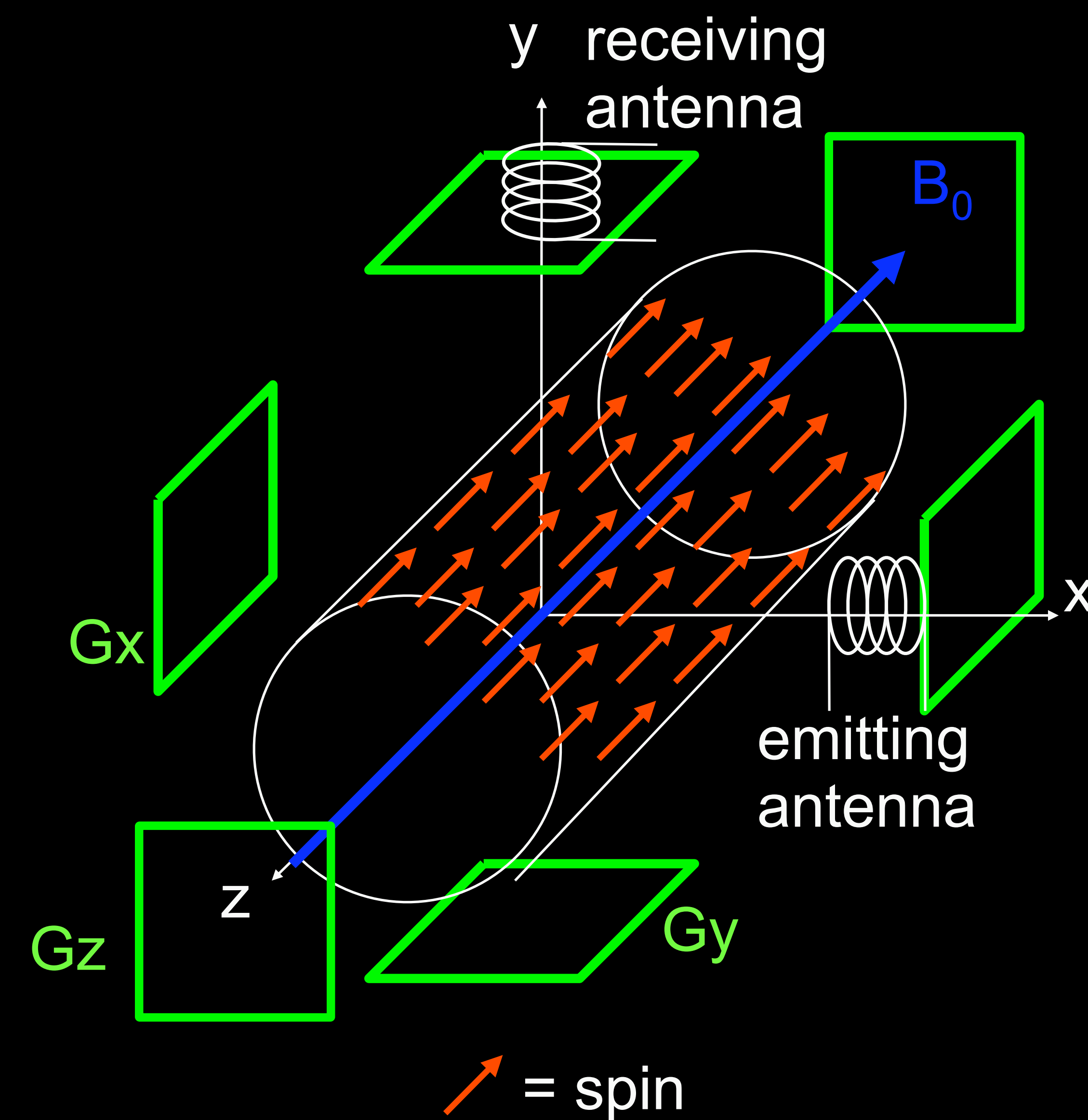


$\omega = \omega_0 + \gamma\, zG_z$ so : $dz = d\omega/(\gamma\, G_z)$

Strong gradient leads to small thickness

MR sequence



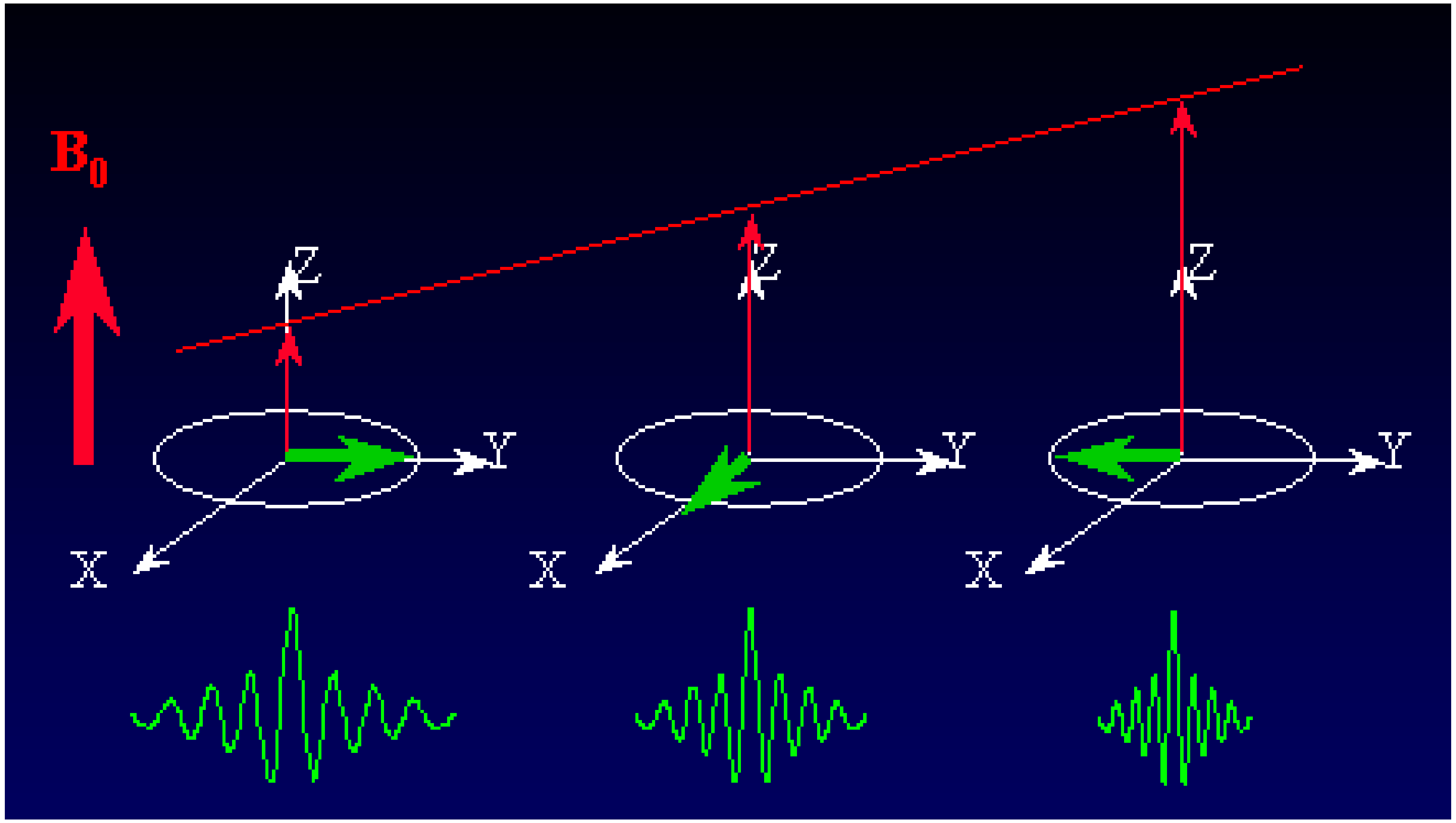


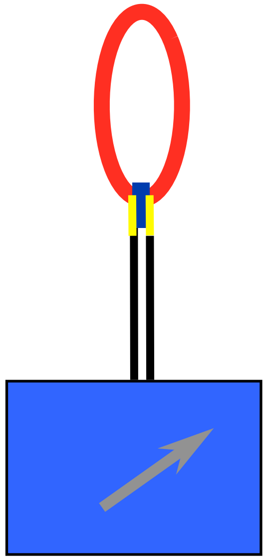
Sequence = 3 gradients (Gx, Gy, Gz)
+ 1 radiofrequency (RF) system
+ 1 (or more) acquisition channel (Acq)

1D Frequency Encoding

Gradient along x axis

Precession frequency varies along x axis: $\omega = \gamma(B_0 + xG_x)$



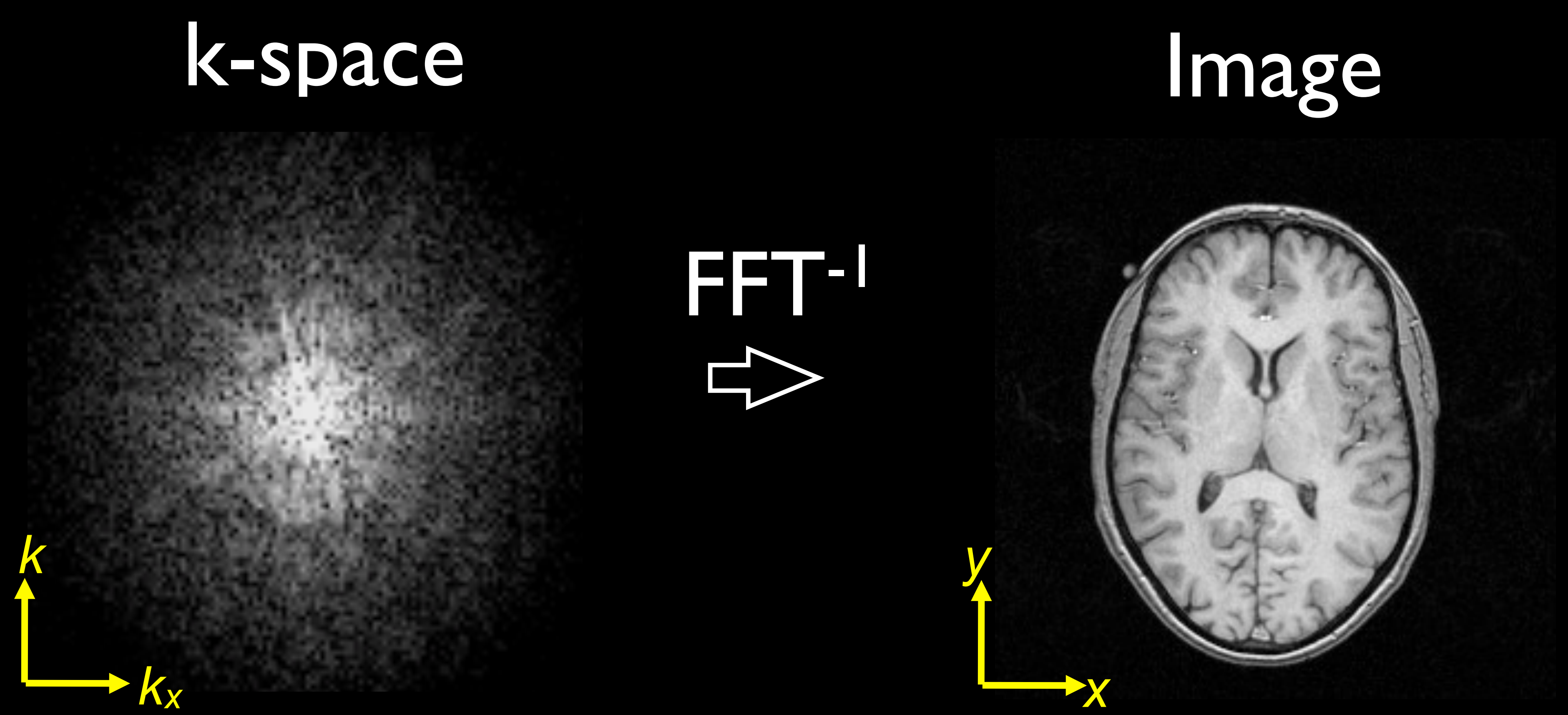


Measured signal: $$S(t) = \sum_y I(y) e^{i\gamma G_y y t}$$

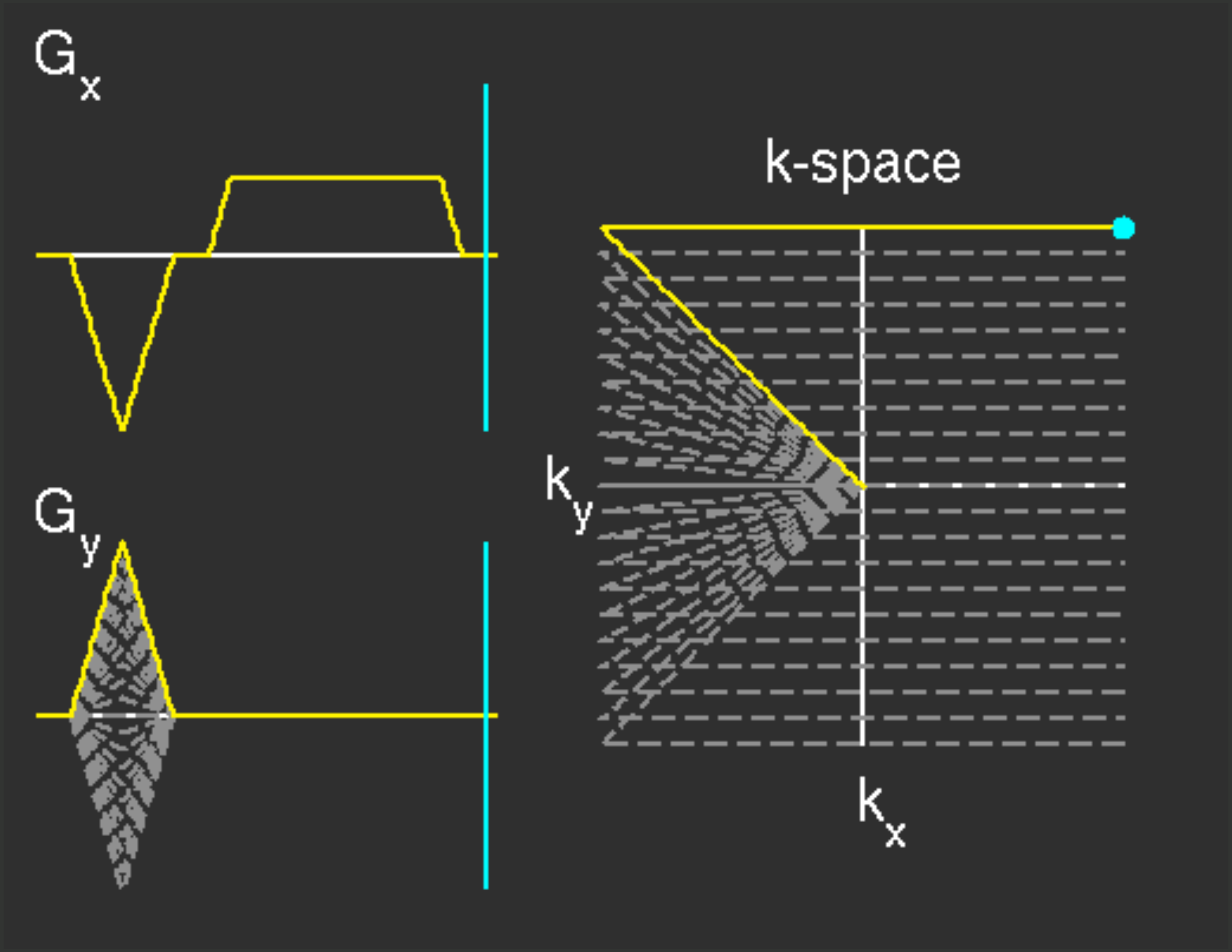
It's a Fourier Transform!

2D Imaging

- Originally obtained by multiple 1D projections by Lauterbur 1973 (like with CT scanner)
- **Phase encoding** is used today instead**:**
  - G_y gradient is used to have a phase that depends on y position
  - Image is obtained with 2D Fourier Transform



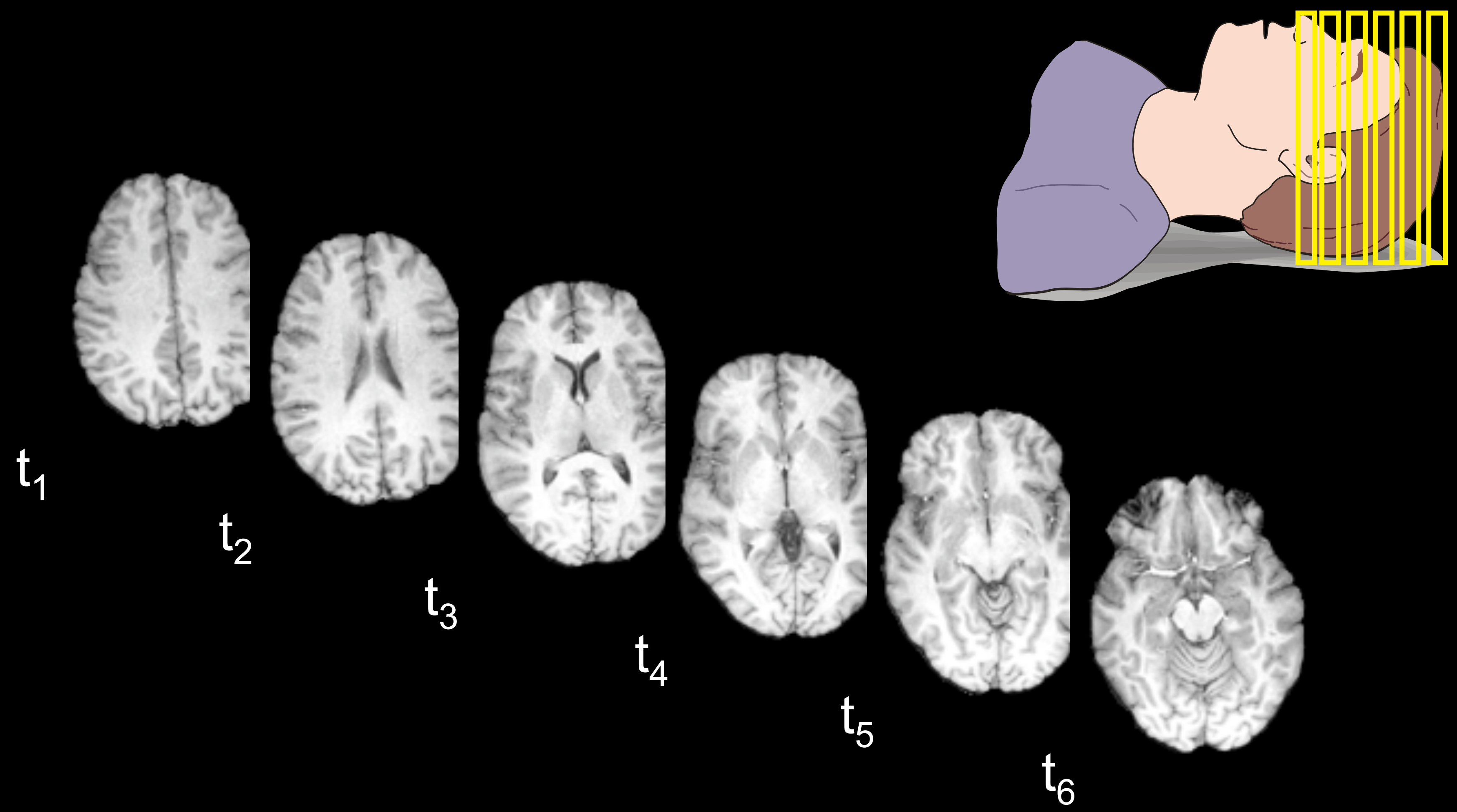
K-space coverage based on accumulated gradient



Gradients are used to cover the k-space
The size and resolution of k-space specifies the ***field of view (FOV)*** (size of image)

If k-space sampling is too low : aliasing problem

Multi-slice imaging



Each slice excited & acquired separately

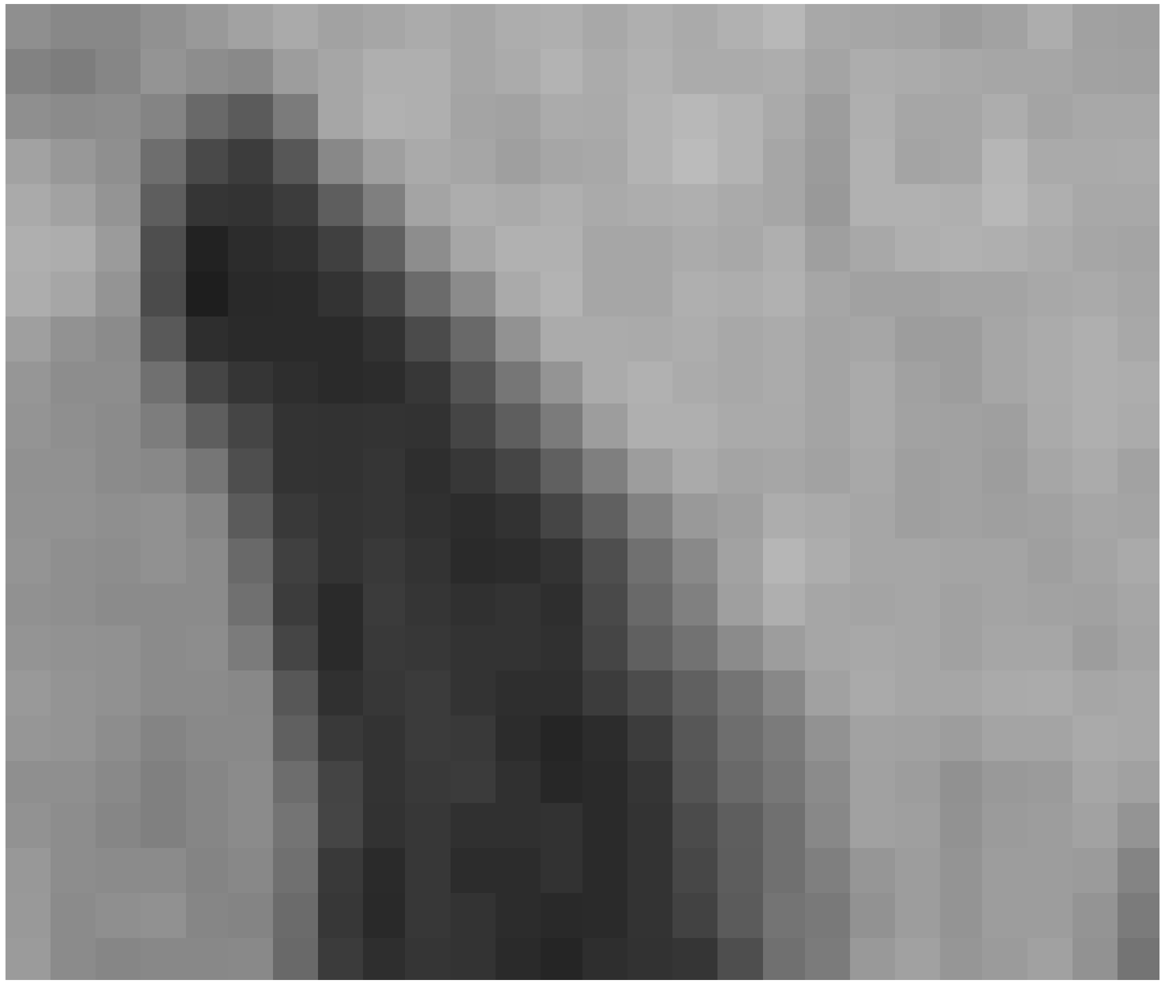
TR: time between repeated excitation of same slice (typically 1–3 seconds)

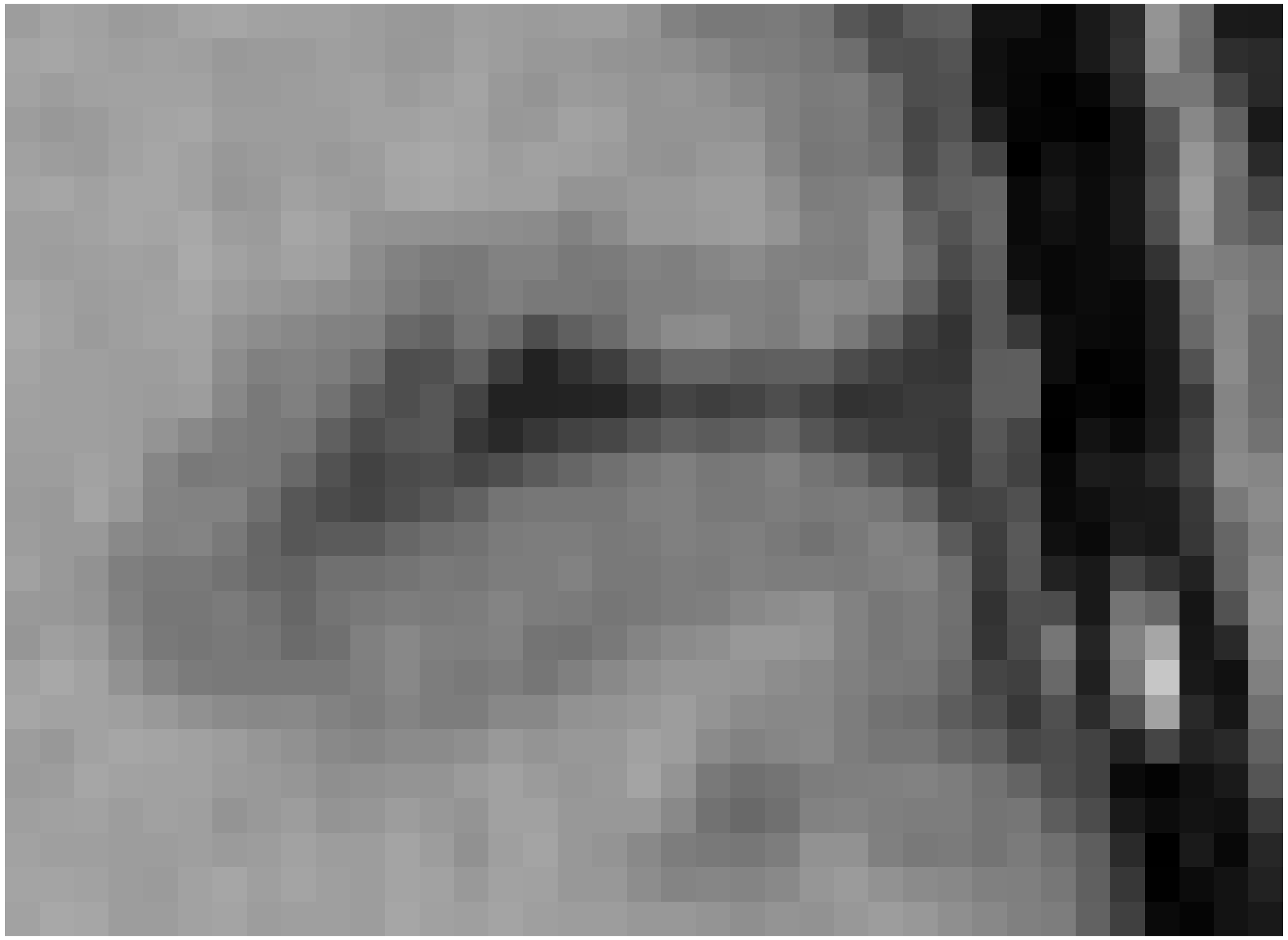
Slices no thinner than ~1 mm

3D imaging

- With multi-slice we need to wait for full relaxation between each slice: ok with long TR like for T2 weighted images
- To accelerate (like for fast T1 weighted images):
  - using symmetry of Fourier plane
  - multi-slice imaging
  - full volume with 2 phase encoding
  - use smaller flip angles
  - Today T1weighted brain scan 8mn. (256x256x256 voxels)

Partial volume effect

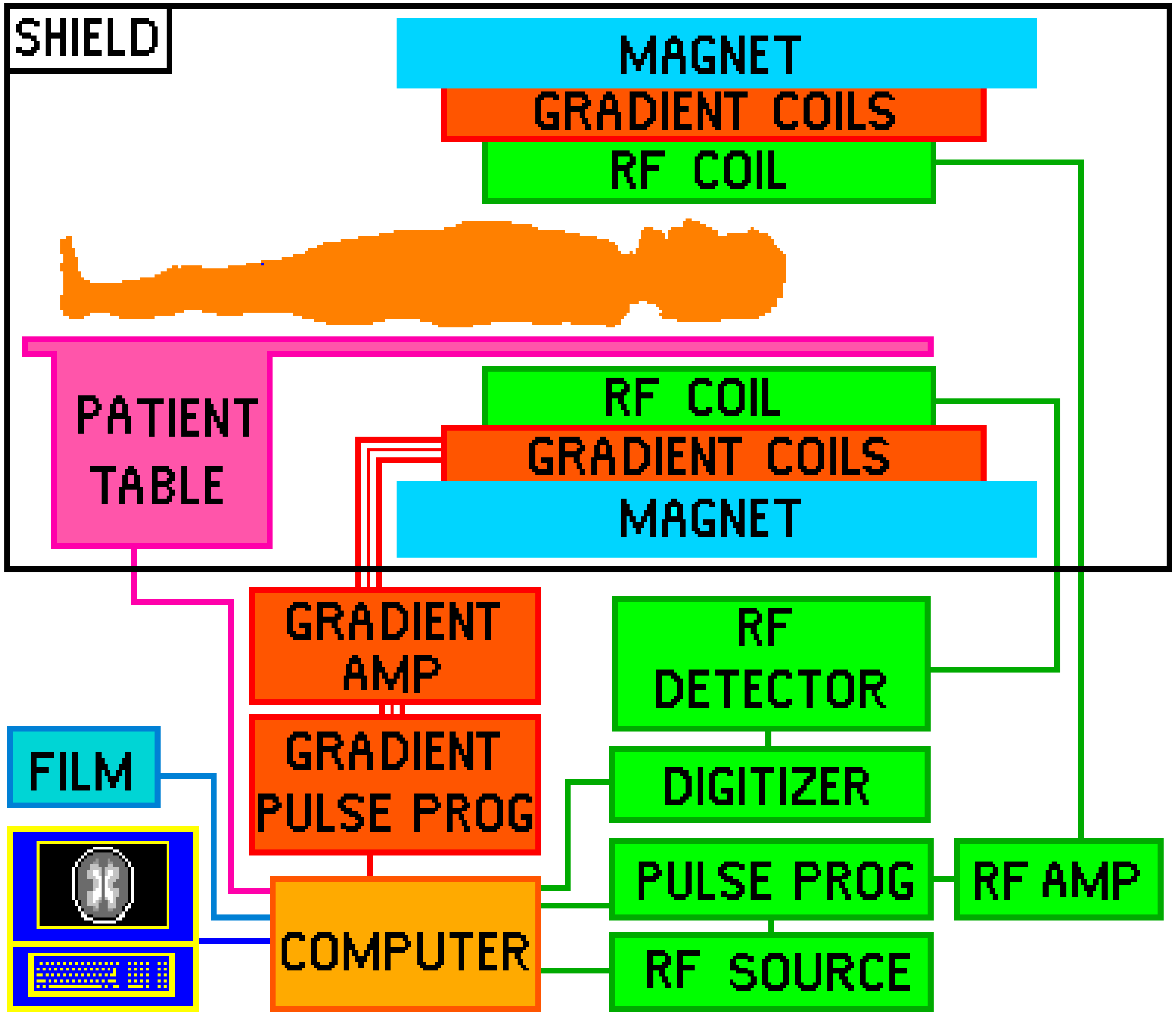




Both WM and GM in 1 voxel

Need smaller voxels

MR system



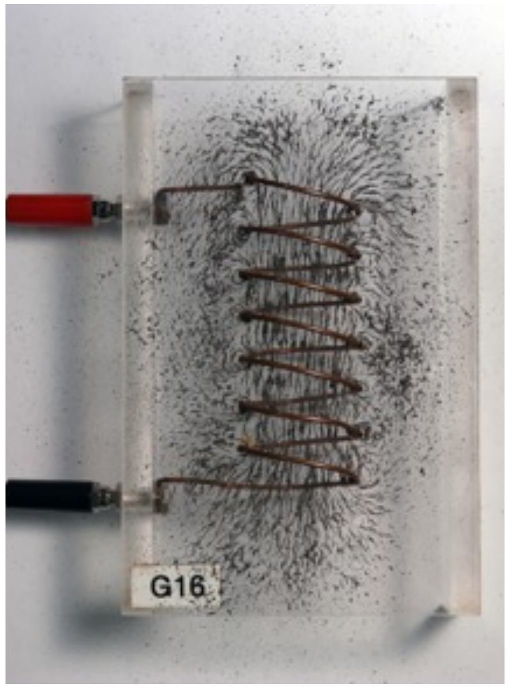
SHIELD
MAGNET
GRADIENT COILS
RF COIL
PATIENT TABLE
RF COIL
GRADIENT COILS
MAGNET
GRADIENT AMP
RF DETECTOR
GRADIENT PULSE PROG
FILM
DIGITIZER
PULSE PROG
RF AMP
COMPUTER
RF SOURCE

RF Fields

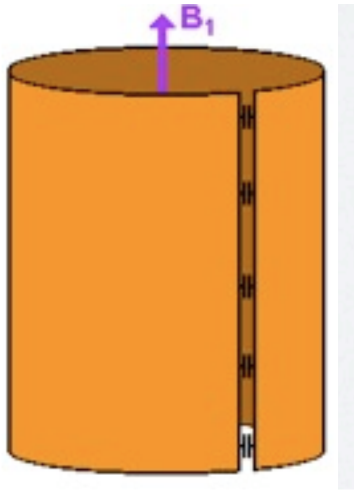
RF fields are generated by RF coils:

Surface

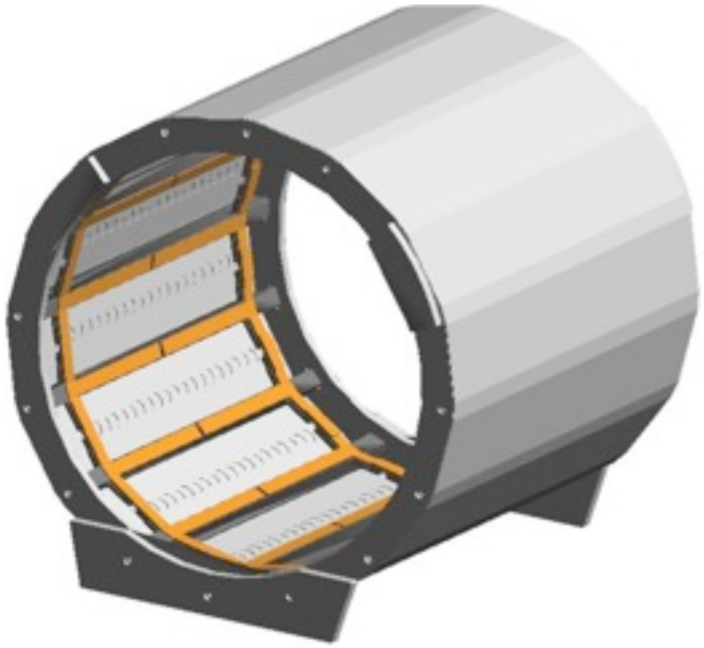
Solenoid



Loop gap



Bird cage coil



- A circularly polarized RF field can be generated using
  - 2 surface coils at right angles: Coils are driven with sinusoidal currents, 90° out of phase "Quadrature" Excitation
  - a bird cage coil

Intrinsic parameters

- T1 : longitudinal relaxation time
- T2 : longitudinal relaxation time
- PD : Proton density
- field inhomogeneity
- physiological motion

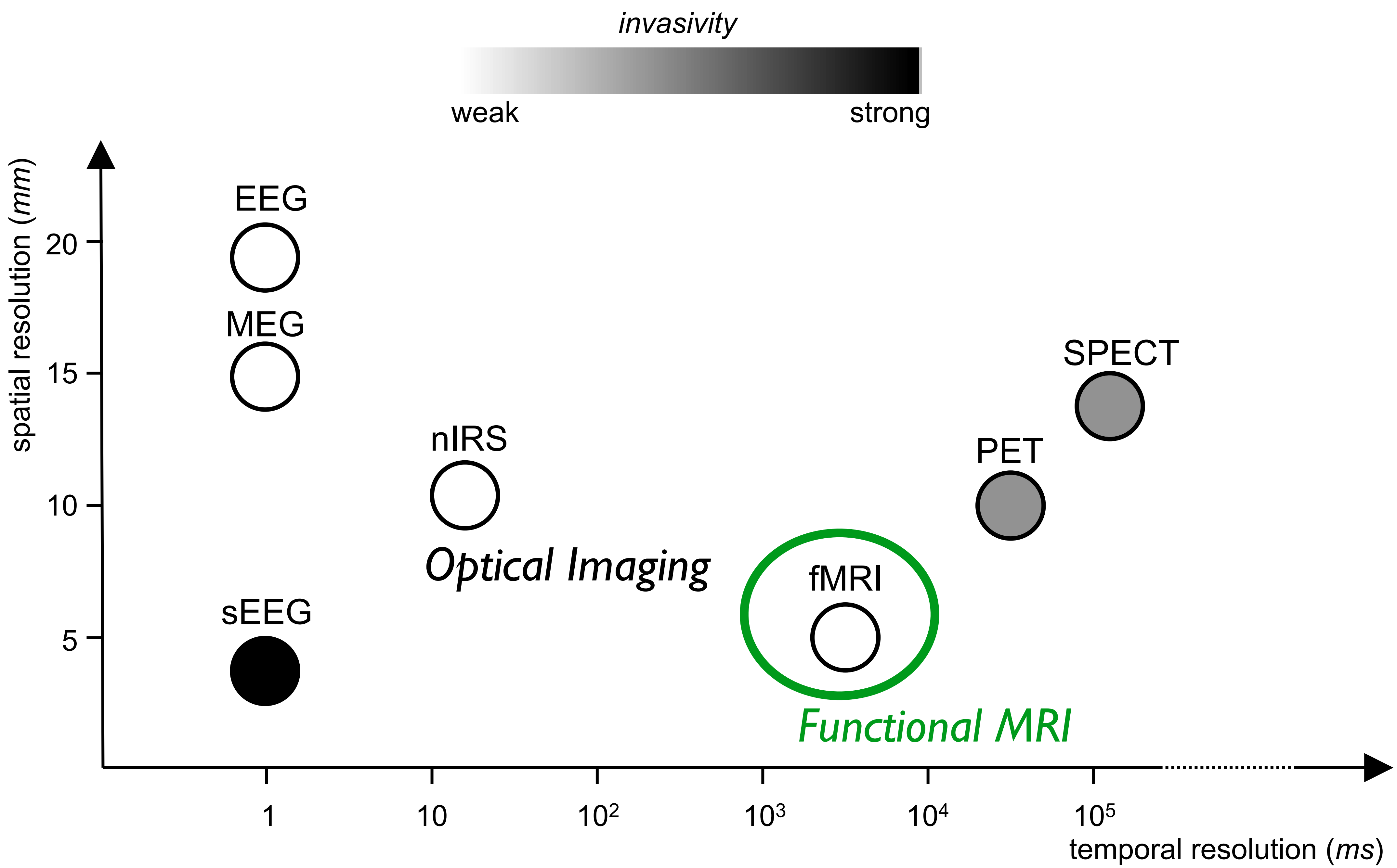
Extrinsic parameters

- External magnetic fields
- Sequence parameters:
  - TR: repetition time
  - TE: echo time (readout)
  - flip angle
  - number of slices
  - FOV
  - slice thickness
  - slice orientation
  - gradient parameters
  - type of coils (surface?)

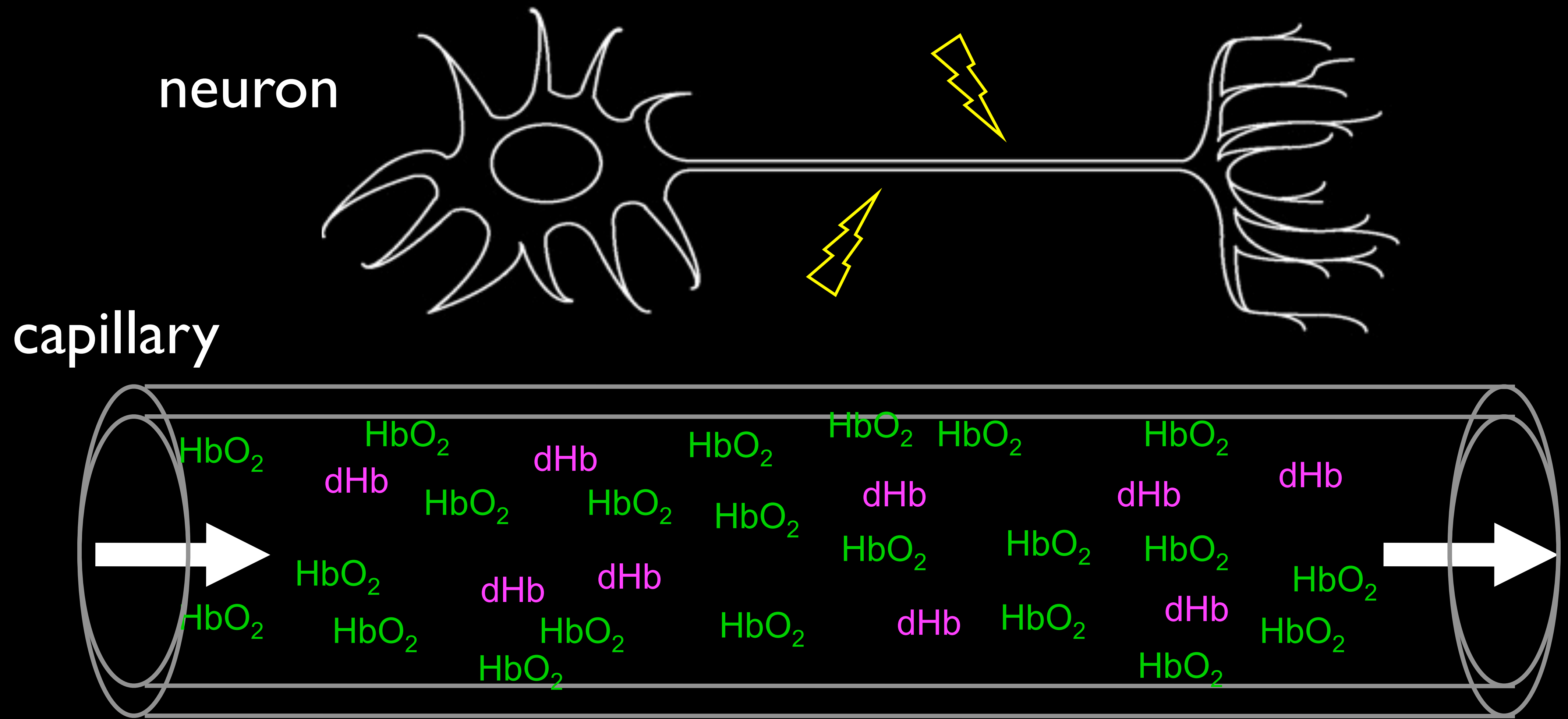
Functional Magnetic Resonance Imaging (fMRI)

Functional neuroimaging

It's the study of the **brain activity** through **functional imaging devices**



Vascular response of activation



↑ O_2 metabolism

↑ blood flow

↑ blood volume

} ↓ [dHb]

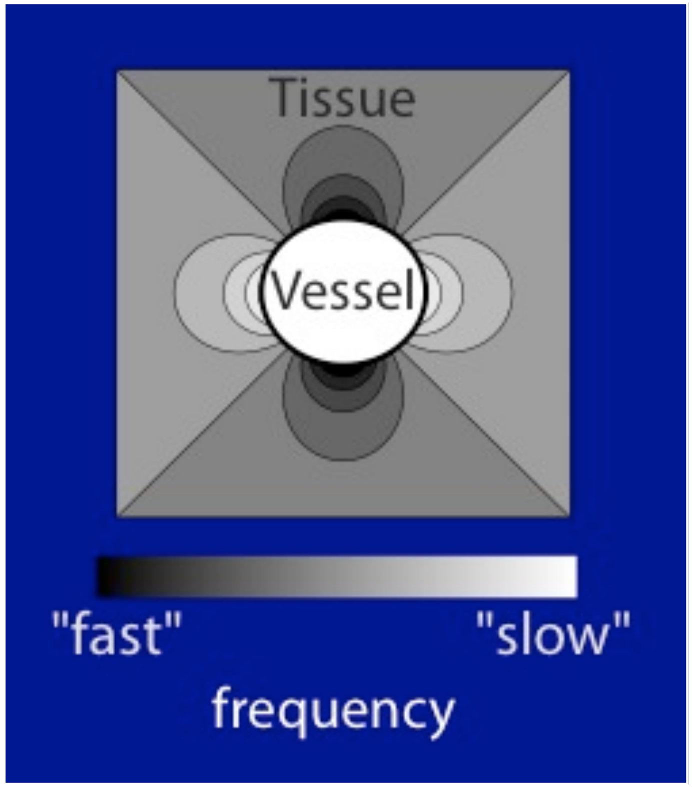
dHb = deoxyhemoglobin

paramagnetic

HbO_2 = oxyhemoglobin

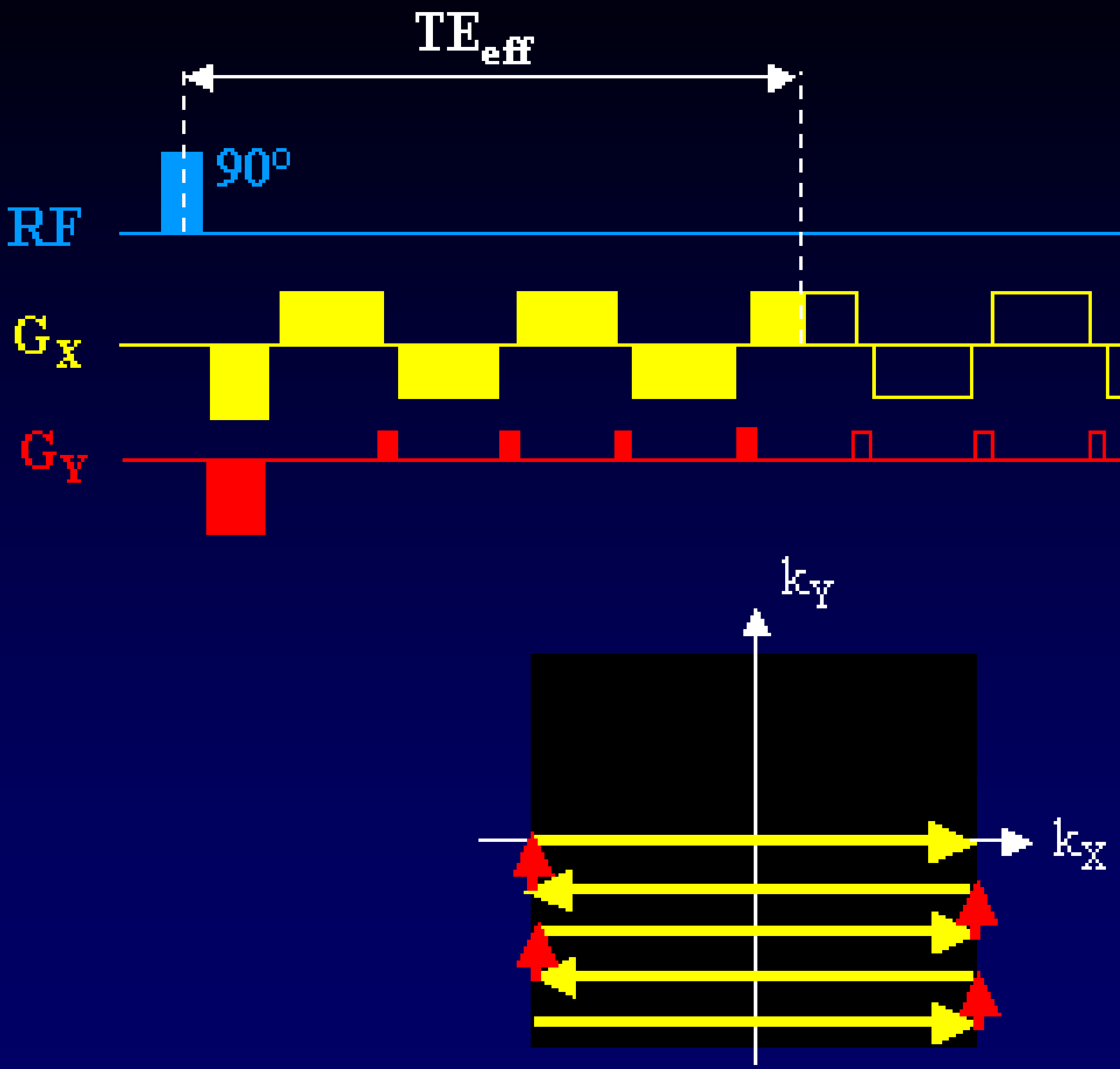
diamagnetic

Functional MRI (fMRI)



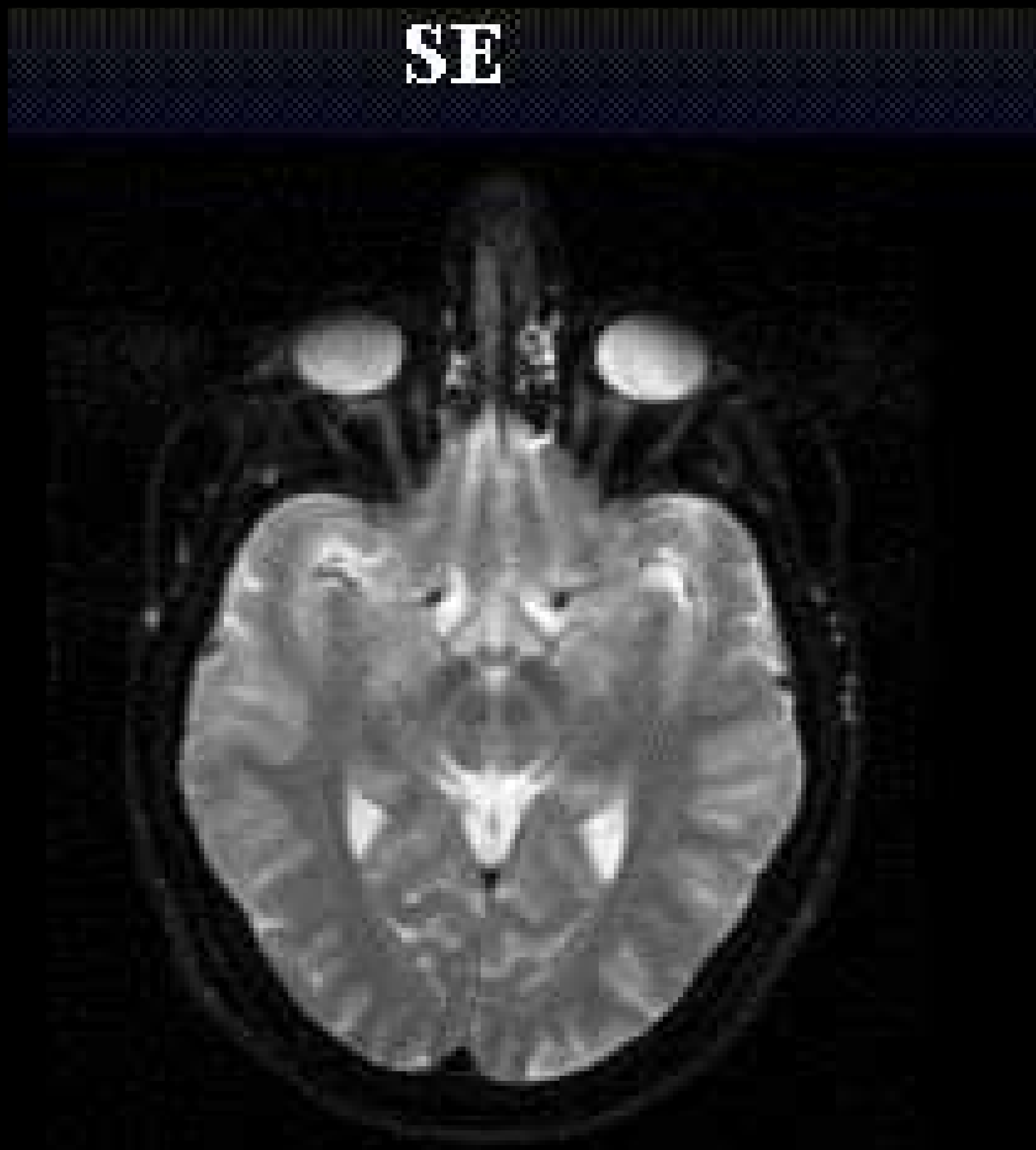
- Blood Oxygenation Level Dependent (BOLD) contrast [Ogawa 1990]
- Deoxyhemoglobin (dHb) has different resonance frequency than water
- dHb acts as endogenous contrast agent: larger T2* detectable with fast imaging sequence
- dHb in blood vessel creates frequency offset in surrounding tissue (approximately as dipole pattern)
- Frequency spread causes signal loss over time
- BOLD contrast: Amount of signal loss reflects [dHb]
- Contrast increases with delay (TE = echo time)

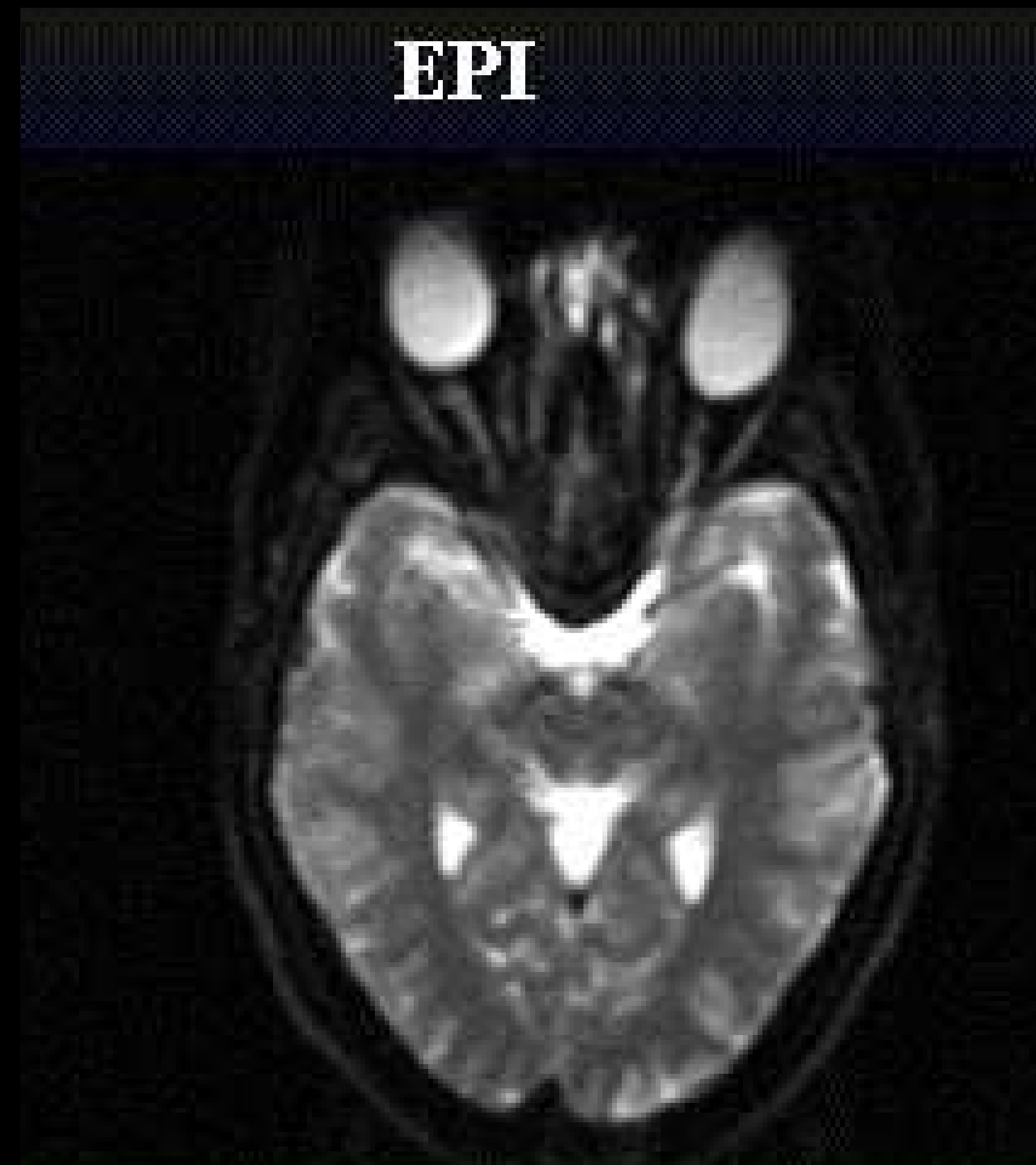
Echo planar imaging (EPI) sequence



Need fast gradients!

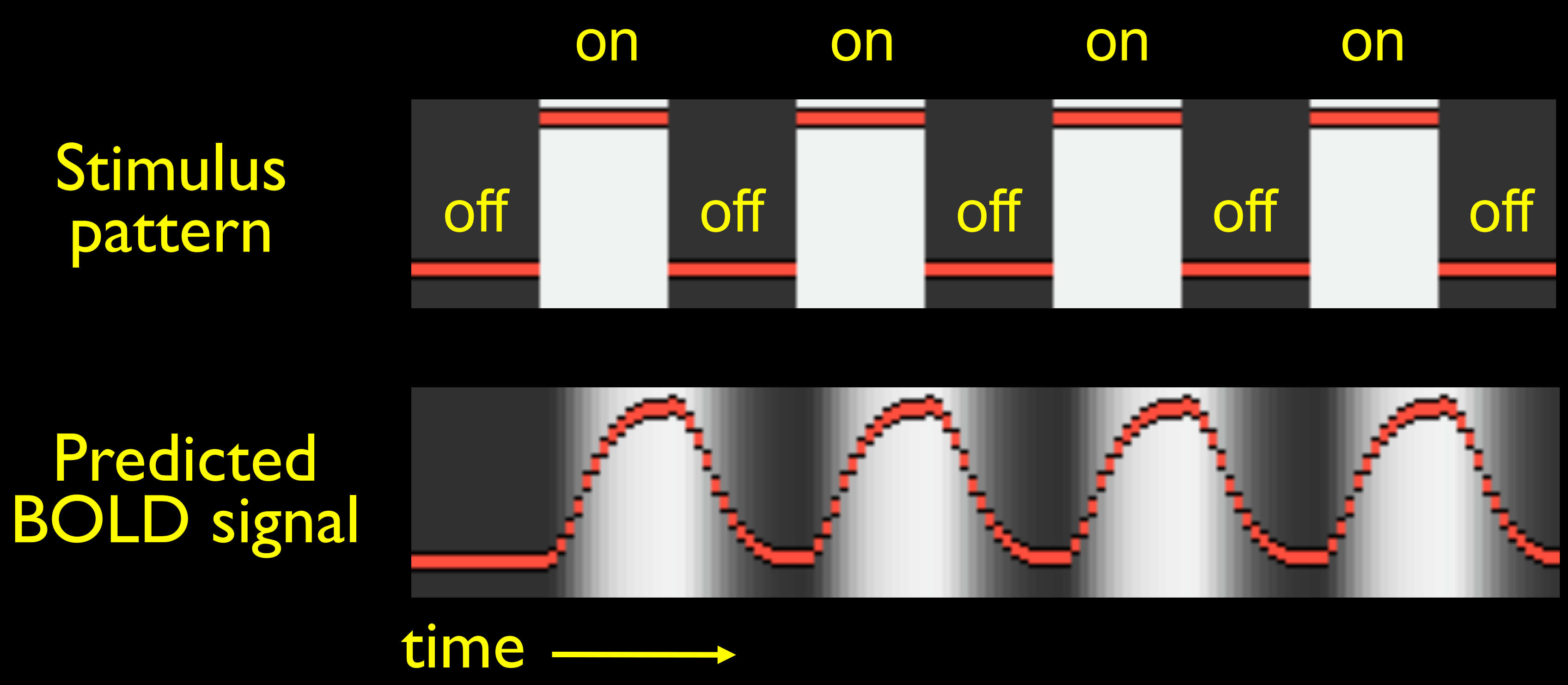
EPI distortions





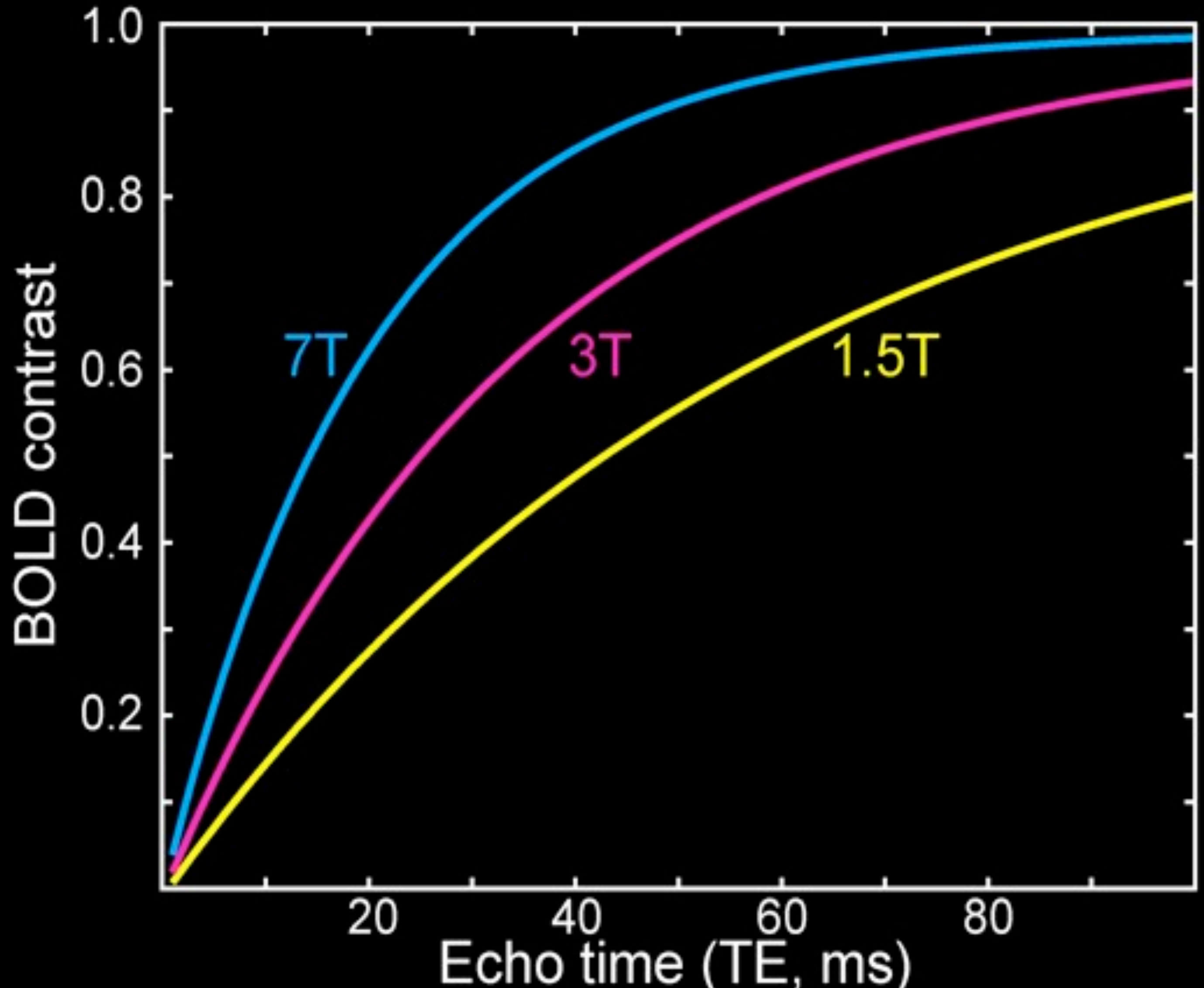
Magnetization precesses at a different rate than expected
Reconstruction places the signal at the wrong location

Typical fMRI experiment



- Subject is given sensory stimulation or task, interleaved with control or rest condition
- Acquire timeseries of BOLD-sensitive images during stimulation
- Analyze image timeseries to determine where signal changed in response to stimulation

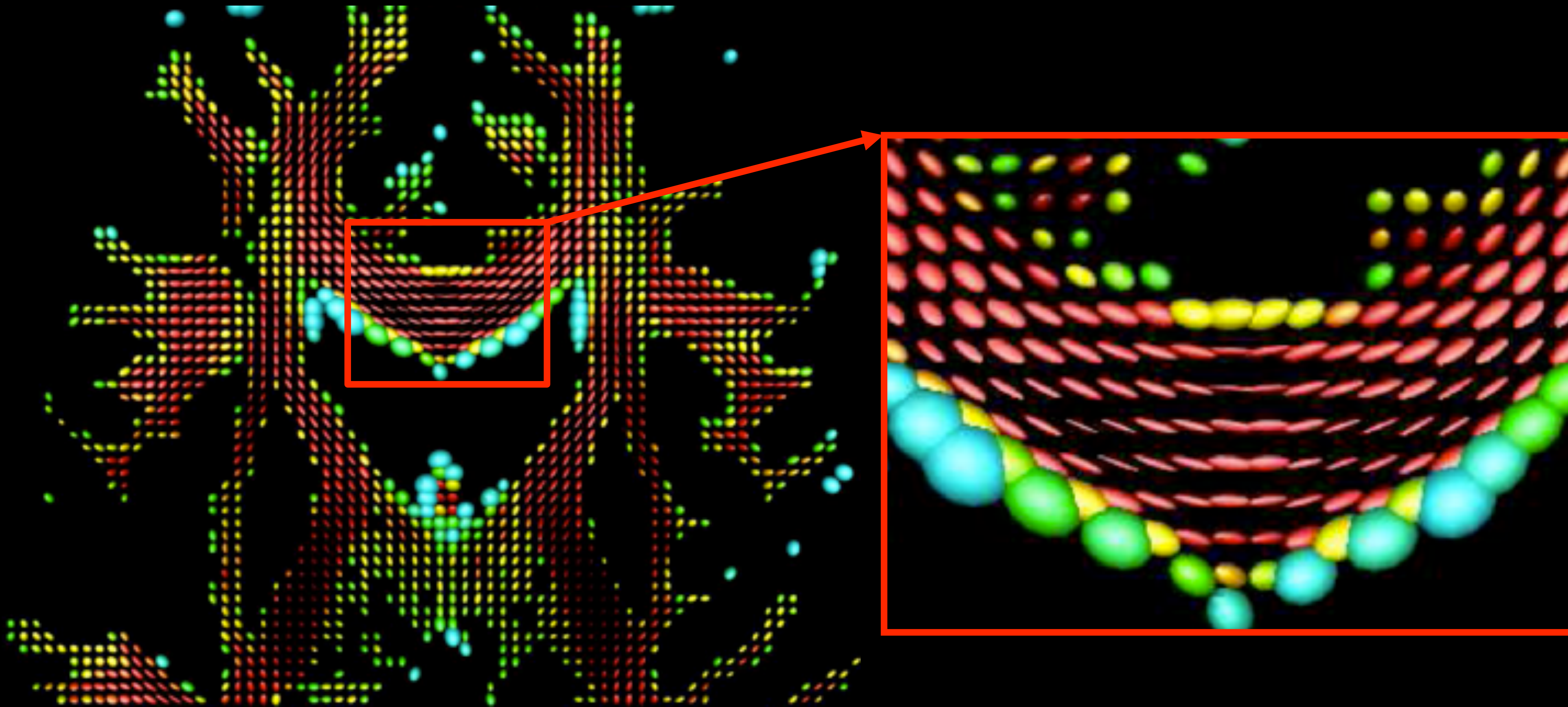
FMRI at High Field (>3T)



- SNR *and* BOLD increase with field strength
- Physiological noise means practical gain is less
- Benefits: Resolution
- Problems: Artifacts, RF heating, wavelength effects…

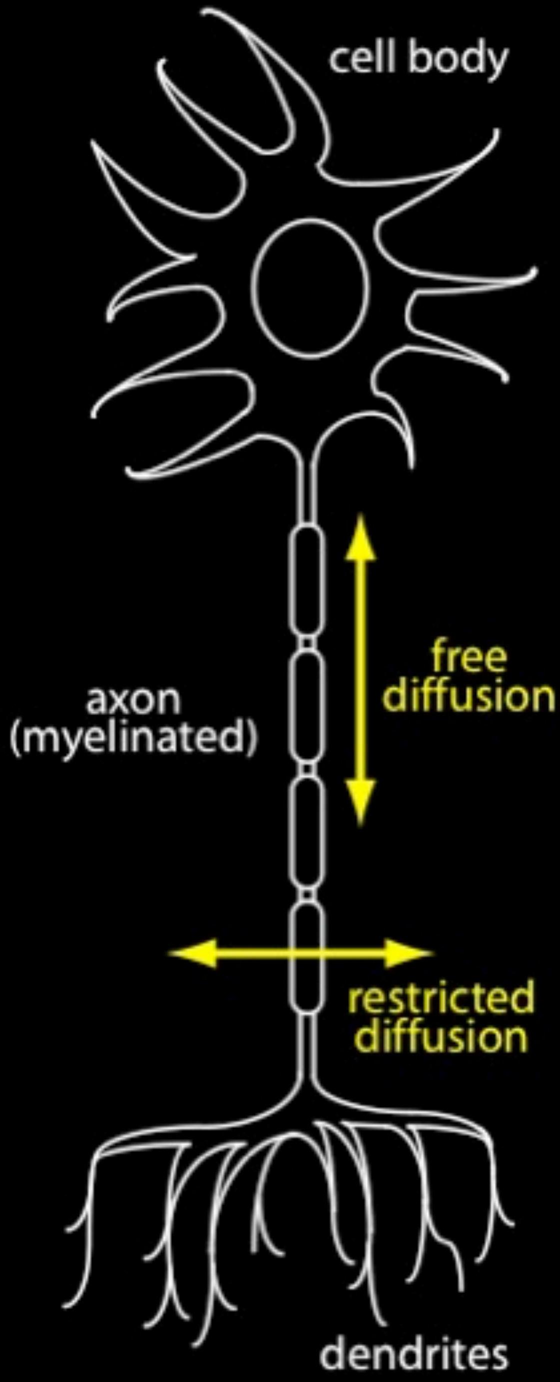
Diffusion Magnetic Resonance Imaging (dMRI)

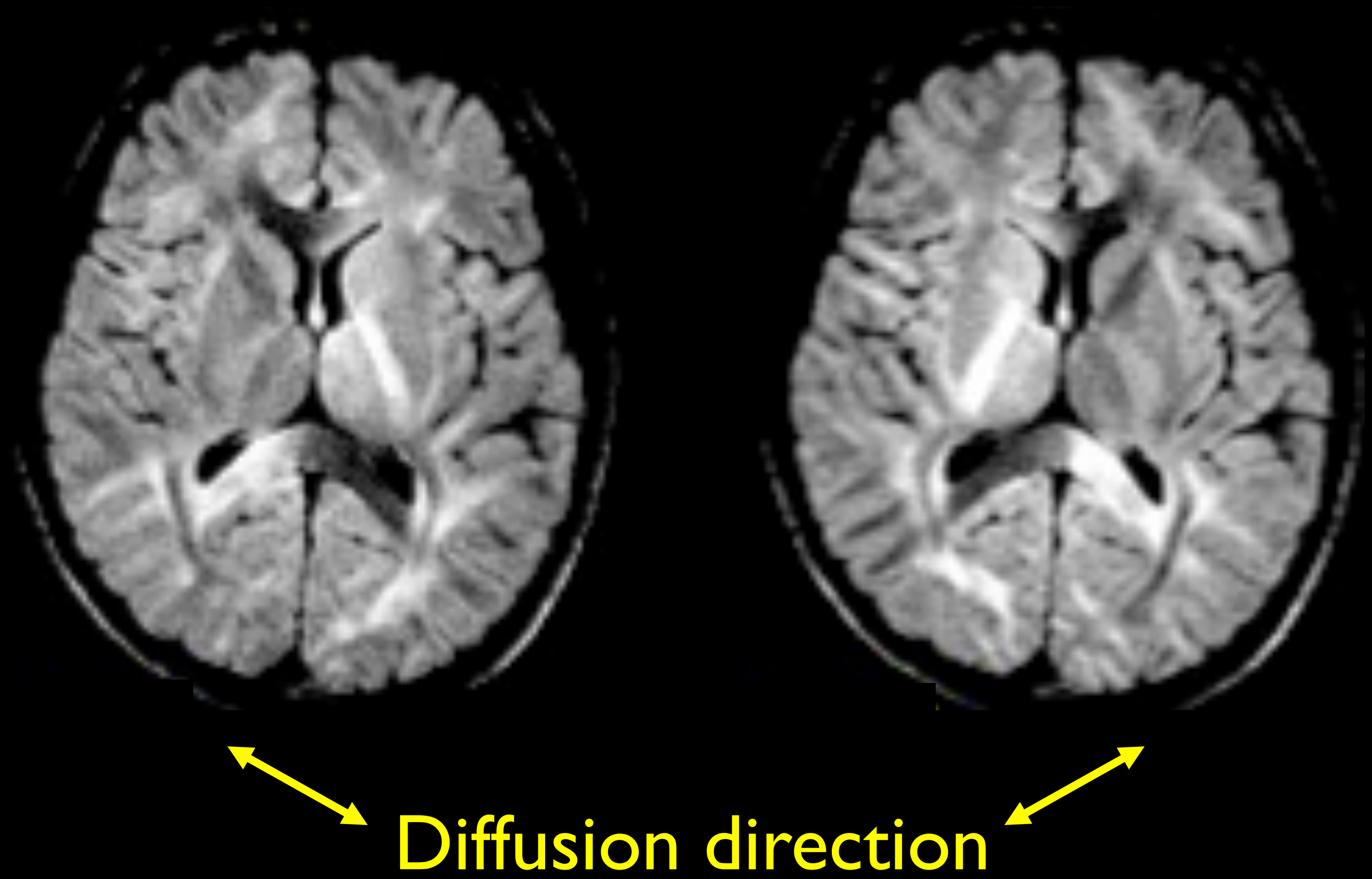
Diffusion MRI (dMRI)



Measures local anisotropy using the signal drop out due to water diffusion

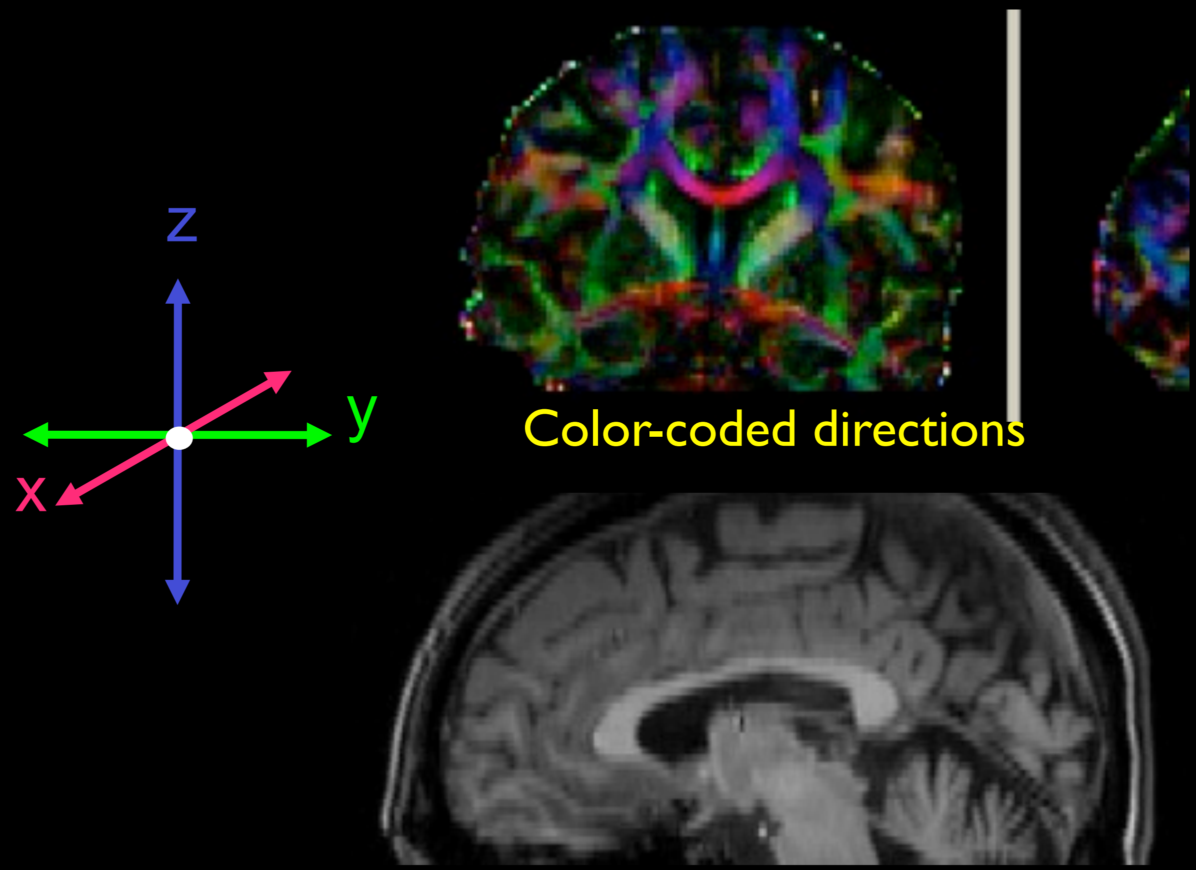
Diffusion MRI (dMRI)

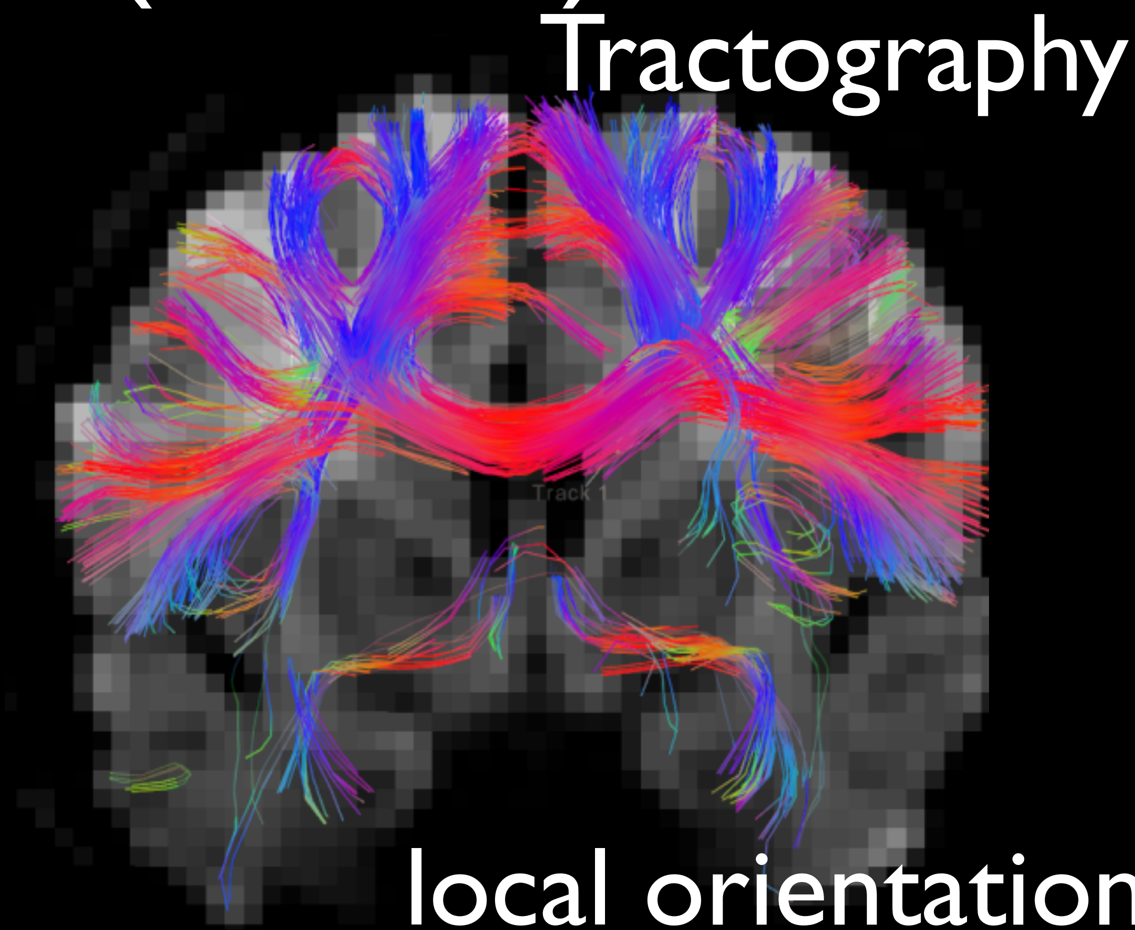


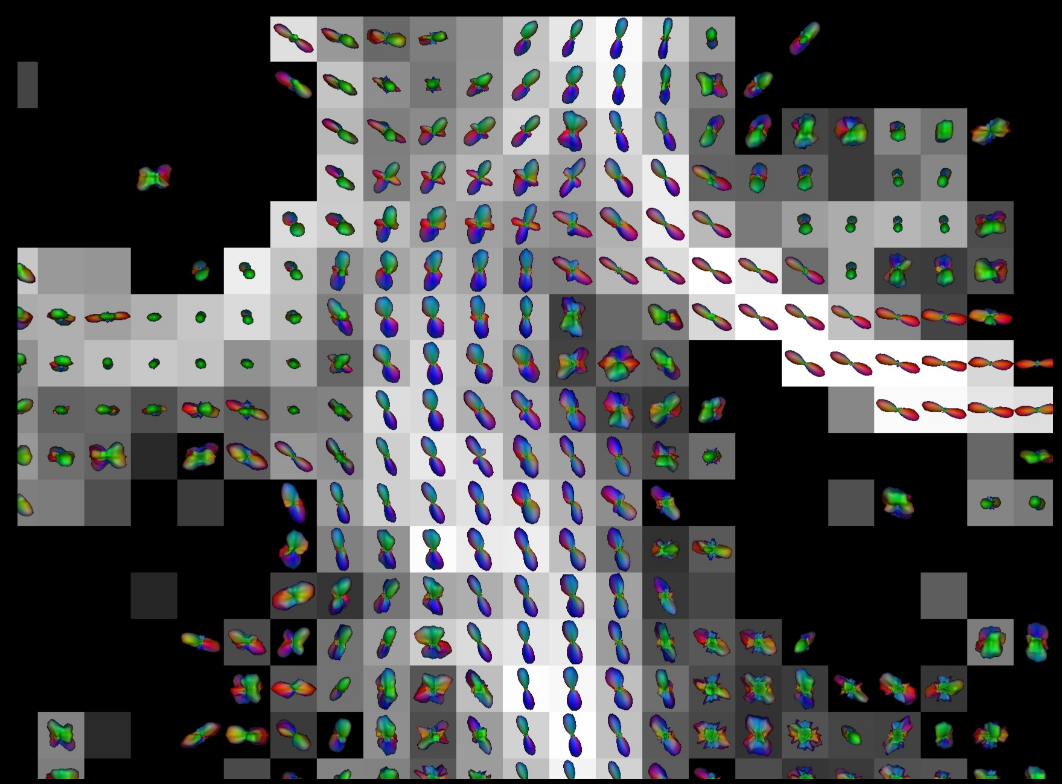


- Water diffusion restricted along white matter
- Sensitize signal to diffusion in different directions
- Measure along all directions, infer tracts

Diffusion MRI (dMRI)

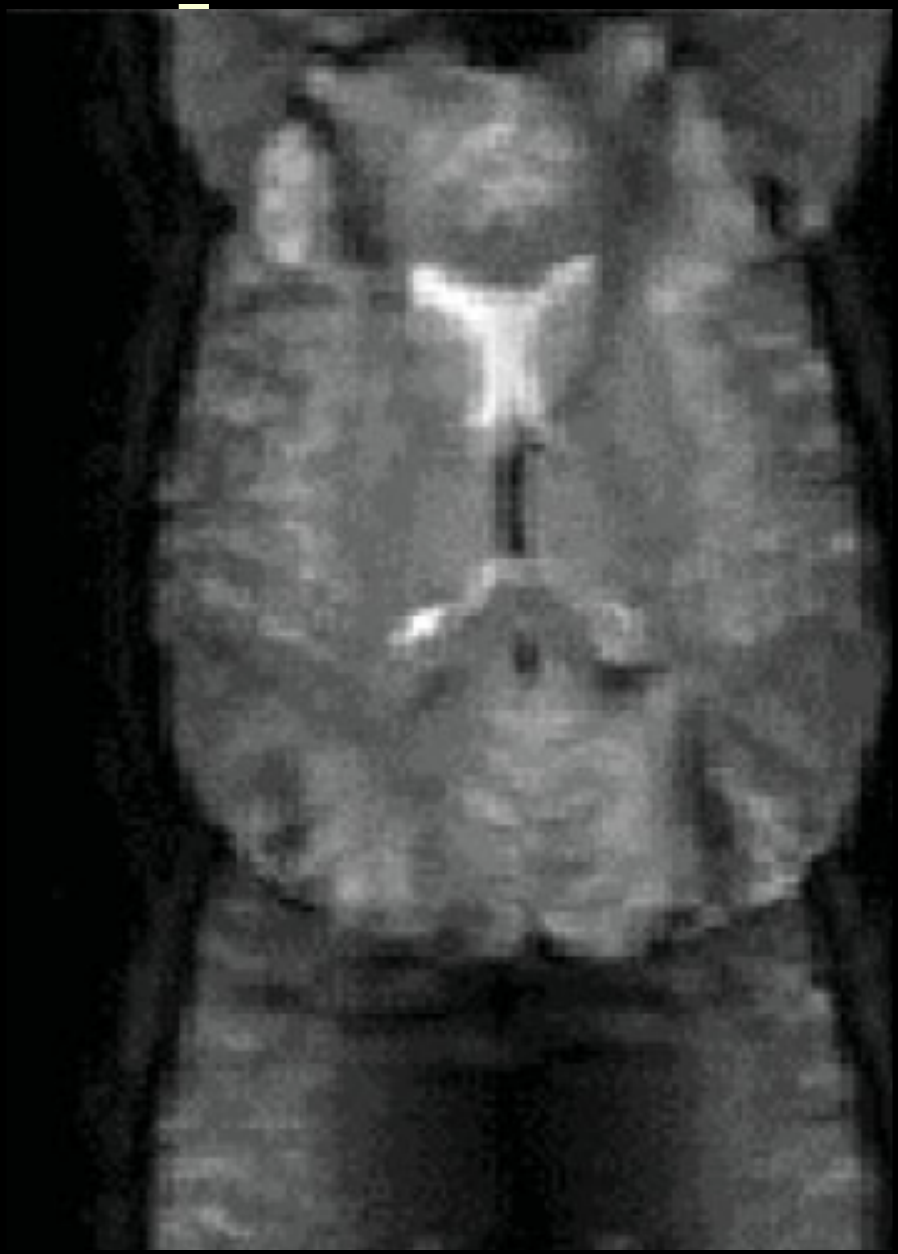






- Complementary information to fMRI
  - fMRI: gray matter, information processing
  - dMRI: white matter, information pathways
- Tractography: tracing white matter pathways between gray matter regions

Artifacts

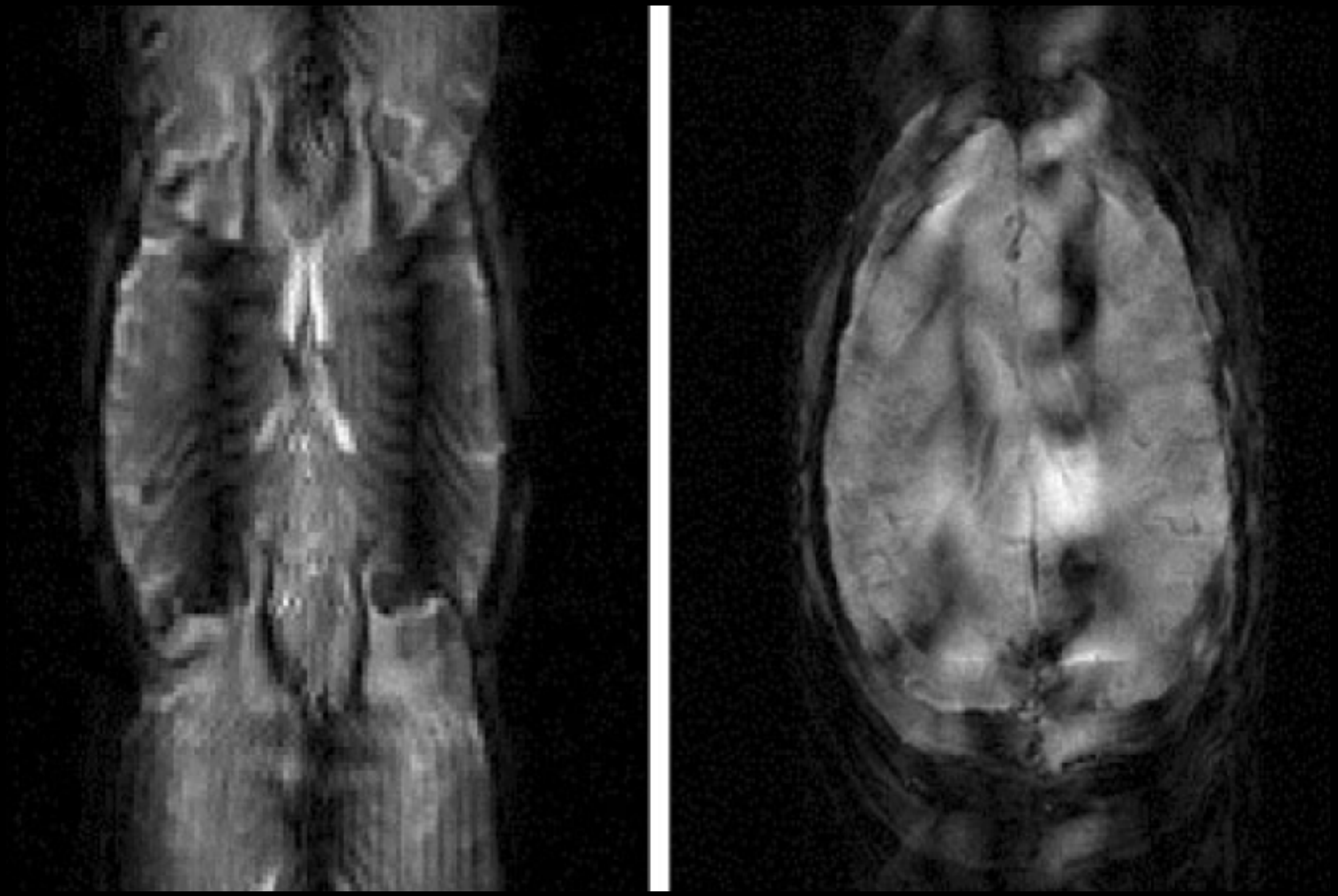


EPI Ghosting

What do you think happened?

Artifacts

What do you think happened?



What else?

MRI with contrast agent (boost signal)
most common Gd-GTPA (paramagnetic)

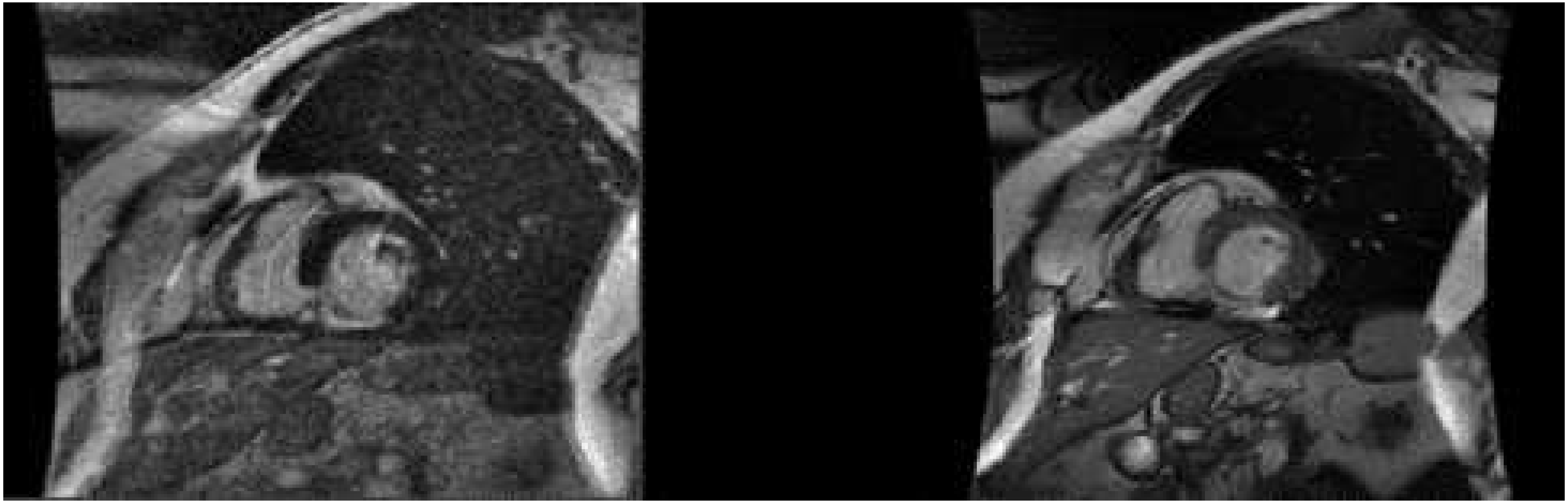
What else?

Angiography

- Image blood flow
- Clinical applications: stenosis, aneurysms

What else?

Cardiac MRI

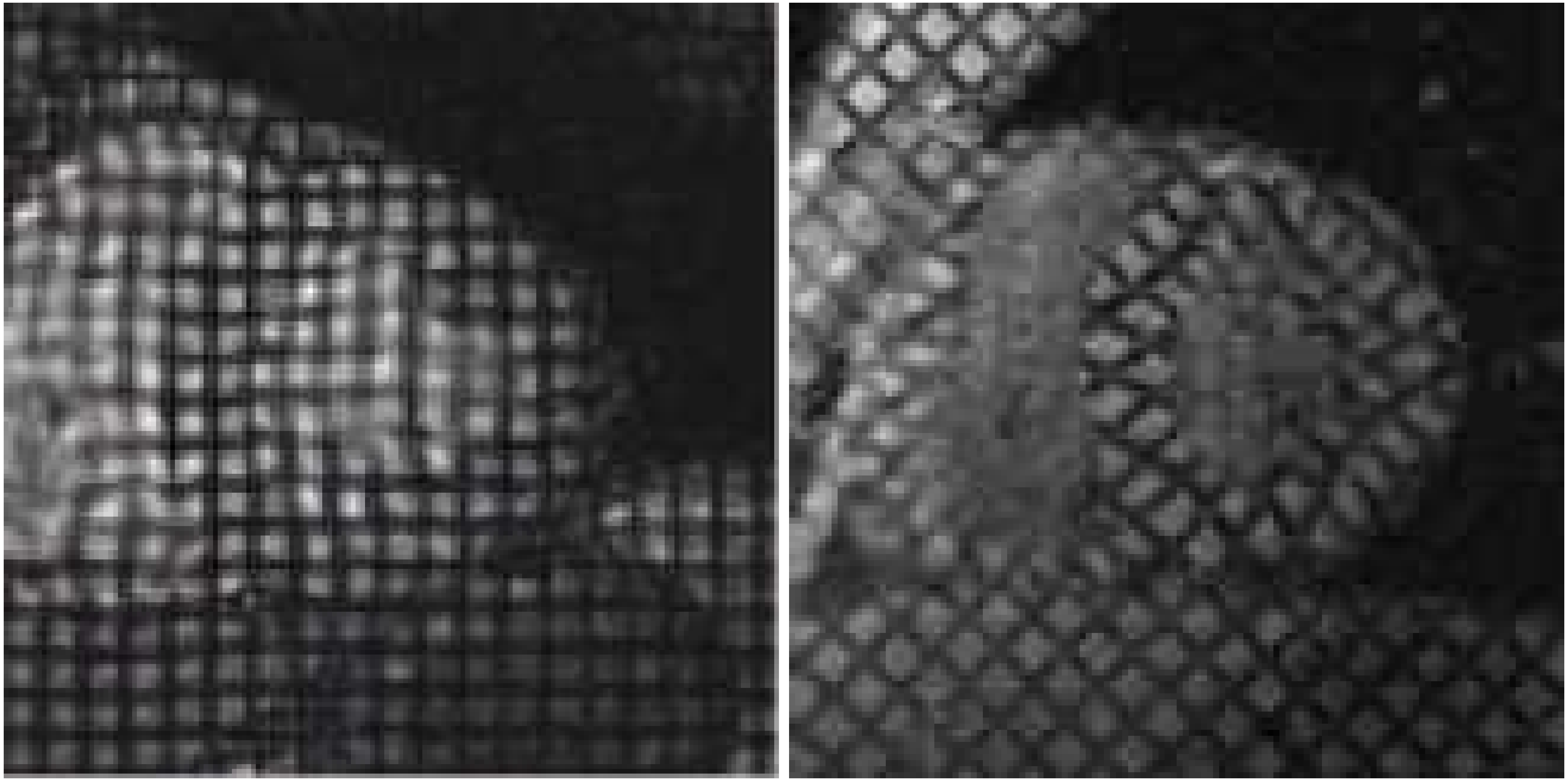


Need to synchronize with heart beat (ECG)

What else?

Tagged MRI

Useful to track tissue motion and distortion



What you should remember

- NMR measures the magnetic properties of nuclei (mainly H)
- Based of precession and electromagnetic induction
- Signal and contrast is obtained by T1 and T2 relaxation
- Intrinsic properties : T1, T2, Proton Density
- Extrinsic : TR, TE, sequence parameters
- Image is obtained by frequency encoding (precession frequency varying with spatial location) and phase encoding.

Learn more

Reading list:

- *Introduction to Functional Magnetic Resonance Imaging*, by R. Buxton
- *The Basics of MRI*, Joseph P. Hornak http://www.cis.rit.edu/htbooks/mri/
- http://www.revisemri.com/
- http://www.mritutor.org/mritutor/
- http://www.ebyte.it/library/Library.html#nmr
- google.com ...