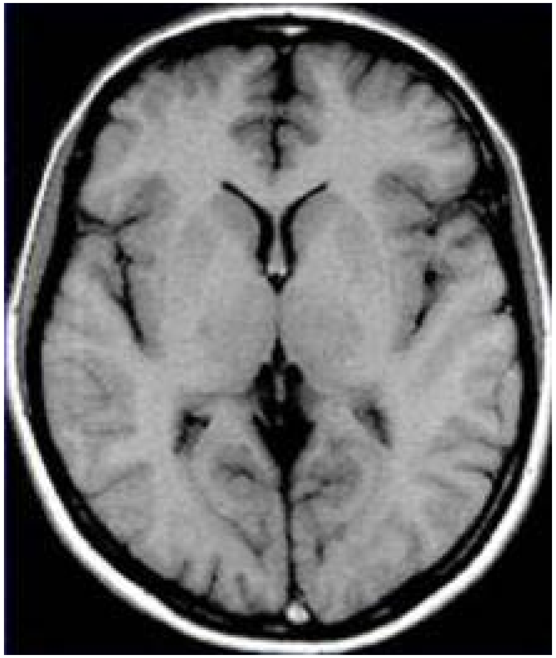

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Master of Science EPF-ETH degree in **Nuclear Engineering**
Medical radiation physics

08. MRI

Introduction

- Limits of imaging techniques using X rays
  - Mostly anatomical
  - Limited contrast of soft tissues

NMR: The basics

Electric charge rotation → magnetic moment μ

$$\mu = I\,A$$

Orbital electron: $\mu = \frac{ev}{2\pi r}\,\pi r^2 = \frac{erv}{2}$

Mass in rotation → kinetic momentum
Spin

$$\vec{L} = \vec{r} \wedge m\vec{v} = J\,\omega$$

$$|\omega| = 2\pi\nu$$

$$L = m_e v r$$

$$\mu = \frac{e}{2m_e} L = \gamma_{orbital}\, L$$

Orbital gyromagnetic ratio

Same phenomenon for electron: electronic gyromagnetic ratio

Same phenomenon for some nucleus: nuclear gyromagnetic ratio

Useful nucleus in NMR

$$^{A}_{Z}X$$

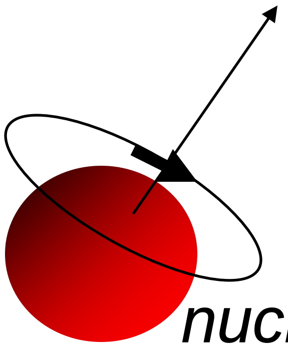
Z atomic number
A mass number

When A and Z even → *no NMR phenomenon*

When A or Z odd → *NMR phenomenon*

When A or Z odd

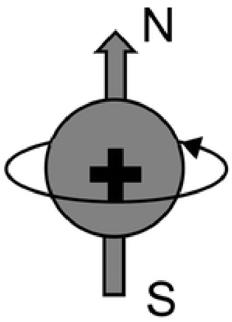
protons, neutrons movement (mass) → quantified kinetic momentum



S = {0, ½ , 1, 3/2 …}

Origin of the signal in MRI: Nucleus

Mass: spin $(\vec{S})$



Charges: magnetic momentum $(\vec{\mu})$

- Rotation of a mass leads to a kinetic momentum
  - Quantified spin
  - Proton, neutron and electron have a spin equal to 1/2

- Rotation of charges leads to a magnetic momentum (small magnet)

$$\vec{\mu} = \gamma \cdot \vec{S}$$

γ : Gyromagnetic ratio

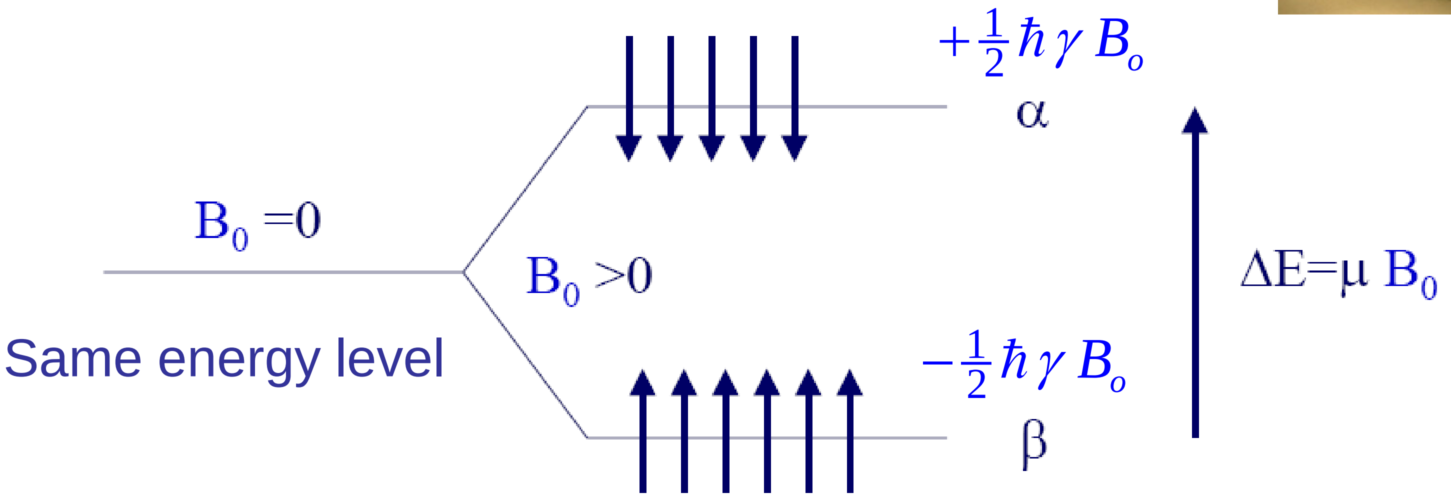
MRI : imaging of hydrogen nuclei → proton imaging

Nucleus properties : spin values

Z / A	Even	Odd
Even	I=0 Ex: ^{12}C, ^{16}O	I = 1 , 2 , 3 , … Ex: ^{14}N, ^{2}H
Odd	I = 1/2, 3/2, 5/2, … Ex: ^{13}C, ^{17}O, ^{1}H , ^{15}N , ^{19}F , ^{31}P	

Effect of a static magnetic field on a magnetic momentum

1. Splitting of energy levels



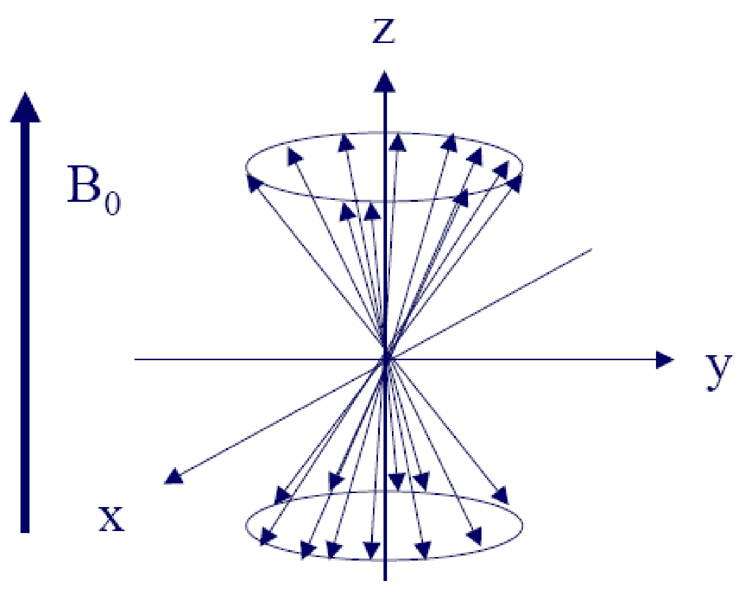
Number of levels available :
2 I + 1

At equilibrium:
$N\beta / N\alpha = \exp - \Delta E / kT$

If B_o = 1.5 T
$N\beta / N\alpha < 1.001$ %

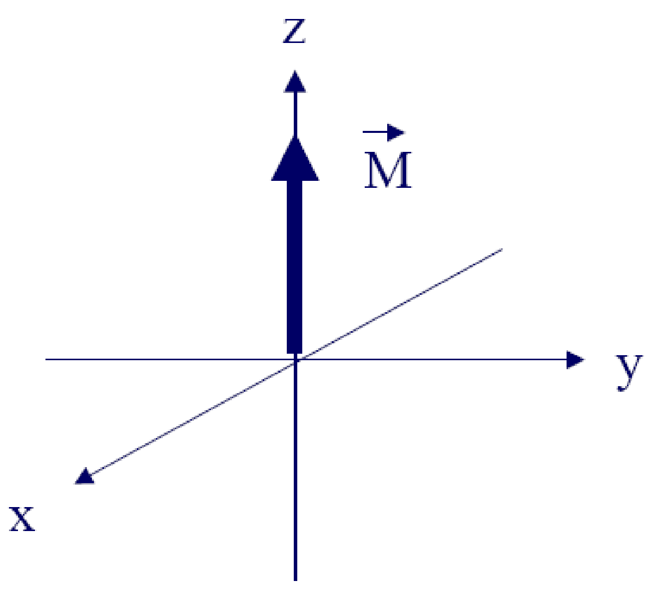
Effect of a static magnetic field on a magnetic momentum

2. A macroscopic magnetization appears



Set of spin ½

Nβ > Nα



$\vec{M}$ = Macroscopic magnetization

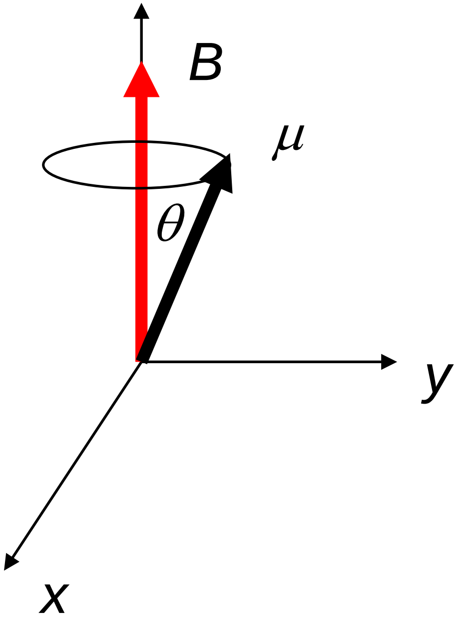
Effect of a static magnetic field on a magnetic momentum

A magnetic field B on a current loop (small magnet or magnetic moment)

→ Torque

$$\vec{C} = \vec{\mu} \wedge \vec{B} \qquad \text{remember}: \vec{F} = \frac{d\vec{p}}{dt} \quad ; \quad \vec{C} = \frac{d\vec{L}}{dt} = \frac{d\vec{I}}{dt} \; ; \; \vec{\mu} = \gamma \, \vec{I}$$

$$\frac{d\vec{I}}{dt} = \vec{\mu} \wedge \vec{B} \Rightarrow \frac{d\,\vec{\mu}}{dt} = \gamma \, \vec{\mu} \wedge \vec{B} \Rightarrow \vec{\omega} = \gamma \, \vec{B} \Leftrightarrow \nu = \frac{\gamma}{2\pi} B$$



→M rotates around B at Larmor frequency

Associated potential energy: $E = \vec{\mu} \bullet \vec{B}$

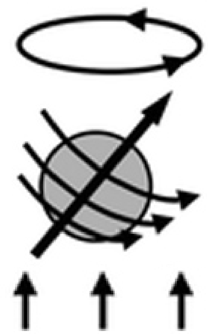
Effect of a static magnetic field on a magnetic momentum

3. Precession of M around B_o

Spinning top in gravitation field

Magnetic moment in a magnetic field (proton)

$$\nu = \frac{\gamma}{2\pi} \cdot \mathbf{B}_o$$



- The spinning frequency is proportional to the static magnetic field B_o (Larmor frequency)
- For proton : ν = 42.58 MHz @ 1 T

Nuclei used in NMR

Nuclei	*S*	*isotopic %.*	*sensibility*	*Larmor frequency (MHz/T)*
H-1	*½*	*99.98*	*1*	*42.576*
H-2	*1*	*0.015*	*0.0096*	*6.535*
P-31	*½*	*100*	*0.0664*	*17.236*
F-19	*½*	*100*	*0.834*	*40.055*
C-13	*½*	*1.108*	*0.0159*	*10.705*
N-15	*½*	*0.365*	*0.00104*	*4.315*
O-17	*5/2*	*0.037*	*0.0291*	*5.772*

(for electron: 176 GHz/T)

NMR or MRI major steps

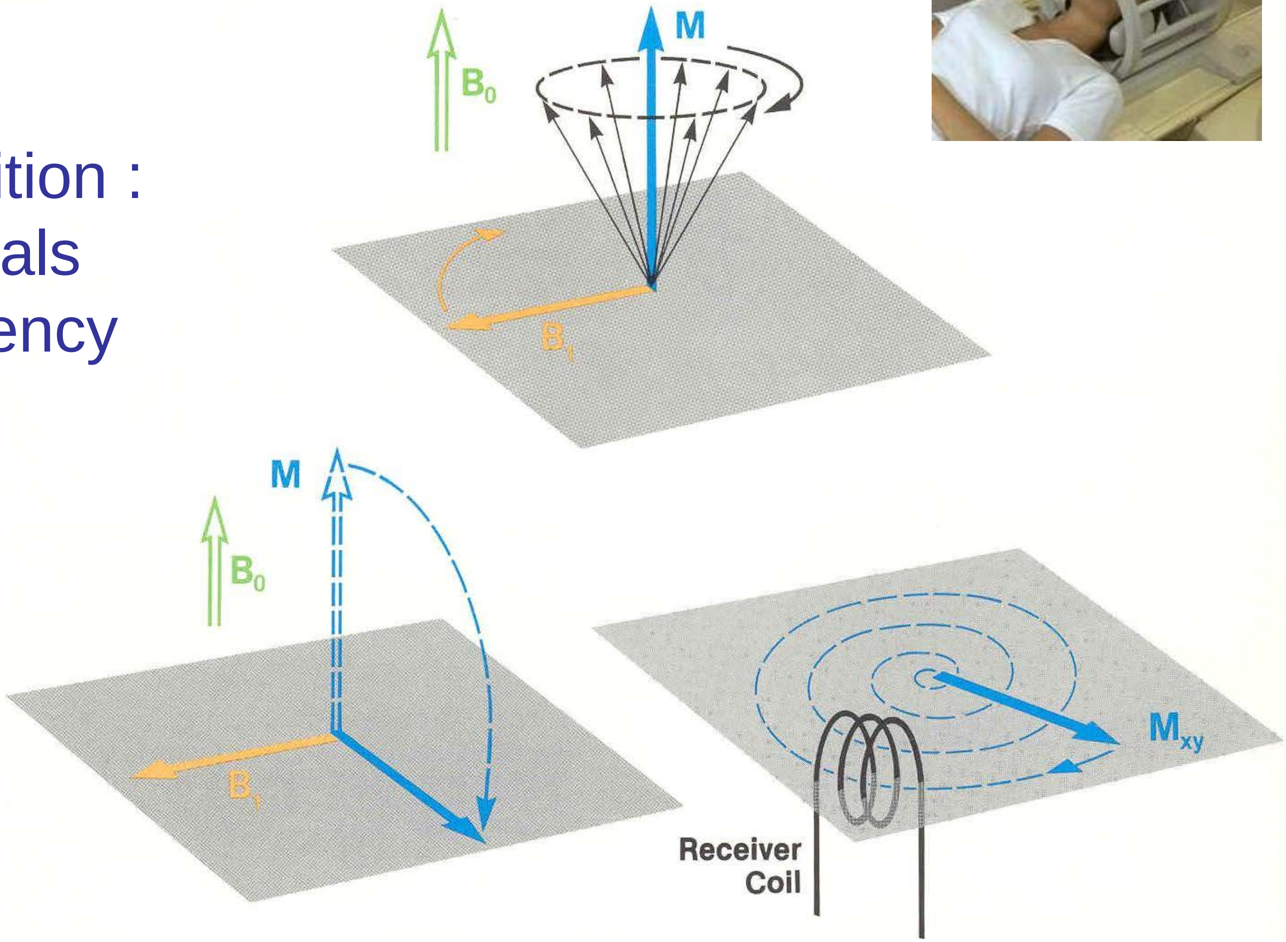
1	2	3
POLARIZATION	*RESONANCE*	*RELAXATION*

Magnetic resonance experiment

A radiofrequency corresponds to a rotating magnetic field B_1

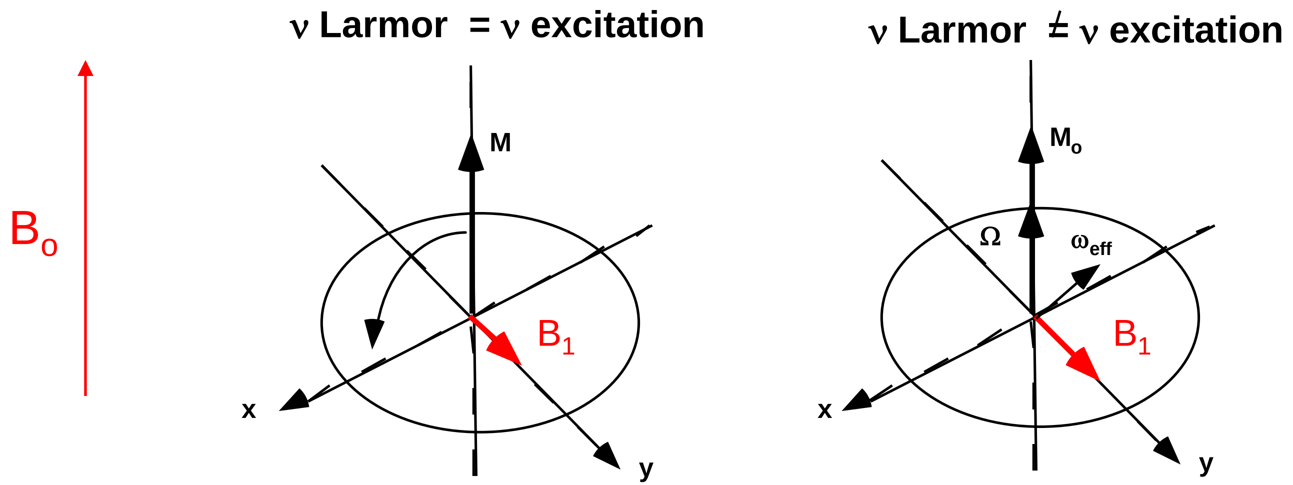
Resonance condition : B_1 frequency equals the Larmor frequency

M rotates around B_0 and B_1



Bloch equation

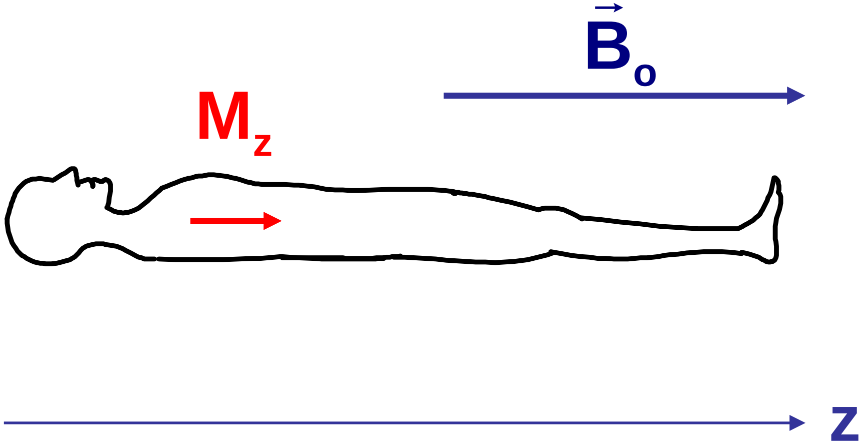
$$\dot{M} = \gamma M \wedge B - R\{M - M_o\} = \begin{cases} \dot{M}_x = \gamma\left[M_y B_z - M_z B_y\right] - M_x / T_2 \\ \dot{M}_y = \gamma\left[M_z B_x - M_x B_z\right] - M_y / T_2 \\ \dot{M}_z = \gamma\left[M_x B_y - M_y B_x\right] - (M_z - M_o) / T_1 \end{cases}$$



$$\alpha = \gamma \int B_1(t)\, dt \quad (radians)$$

Principle of MRI

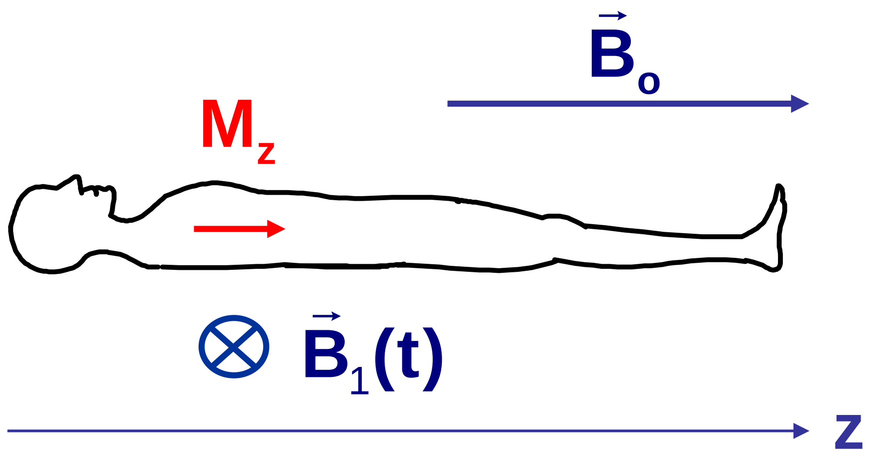
- **The patient is placed in a static magnetic field (B_o)**



B_o creates $\vec{M}$ which projection along z is M_z

Principle of MRI

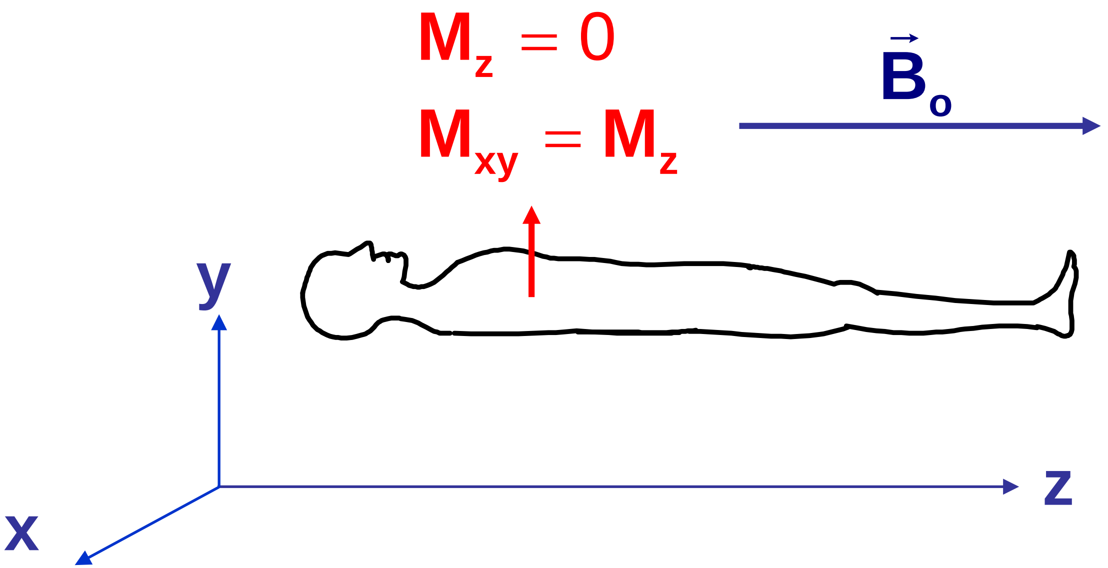
- The direction of M_z is modified by applying a radio frequency, $B_1(t)$, at Larmor frequency (42 MHz/T)
- M_z rotates around B_1 and B_o



Use of an excitation antenna: tilt of M_z in the transverse plane x,y. $M_z \rightarrow M_{xy}$

Principle of MRI

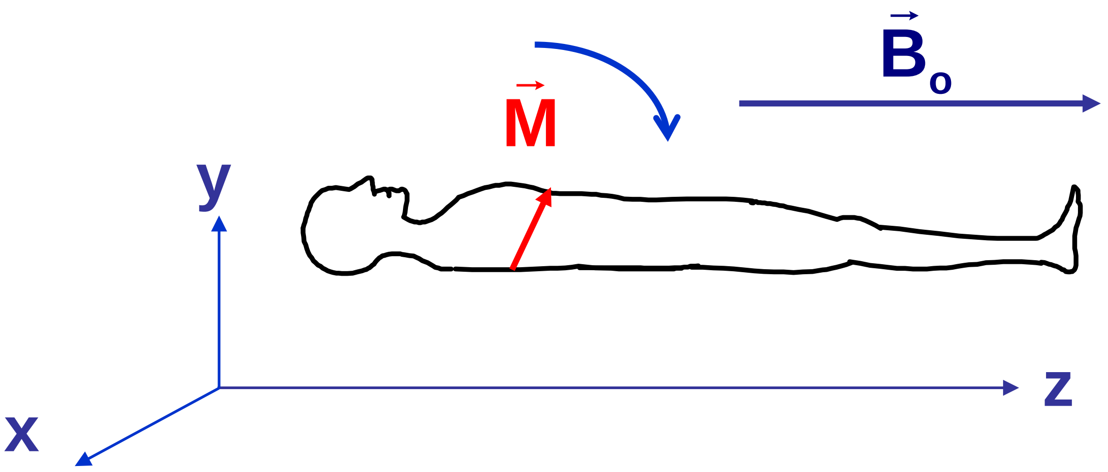
- **For a tilt of 90° M_z is now equal to 0**



The magnetization vector is now in the transverse plane and rotates around B_o (B_1 is switched off). One can detect M_{xy} in that plane with a reception antenna.

Principle of MRI

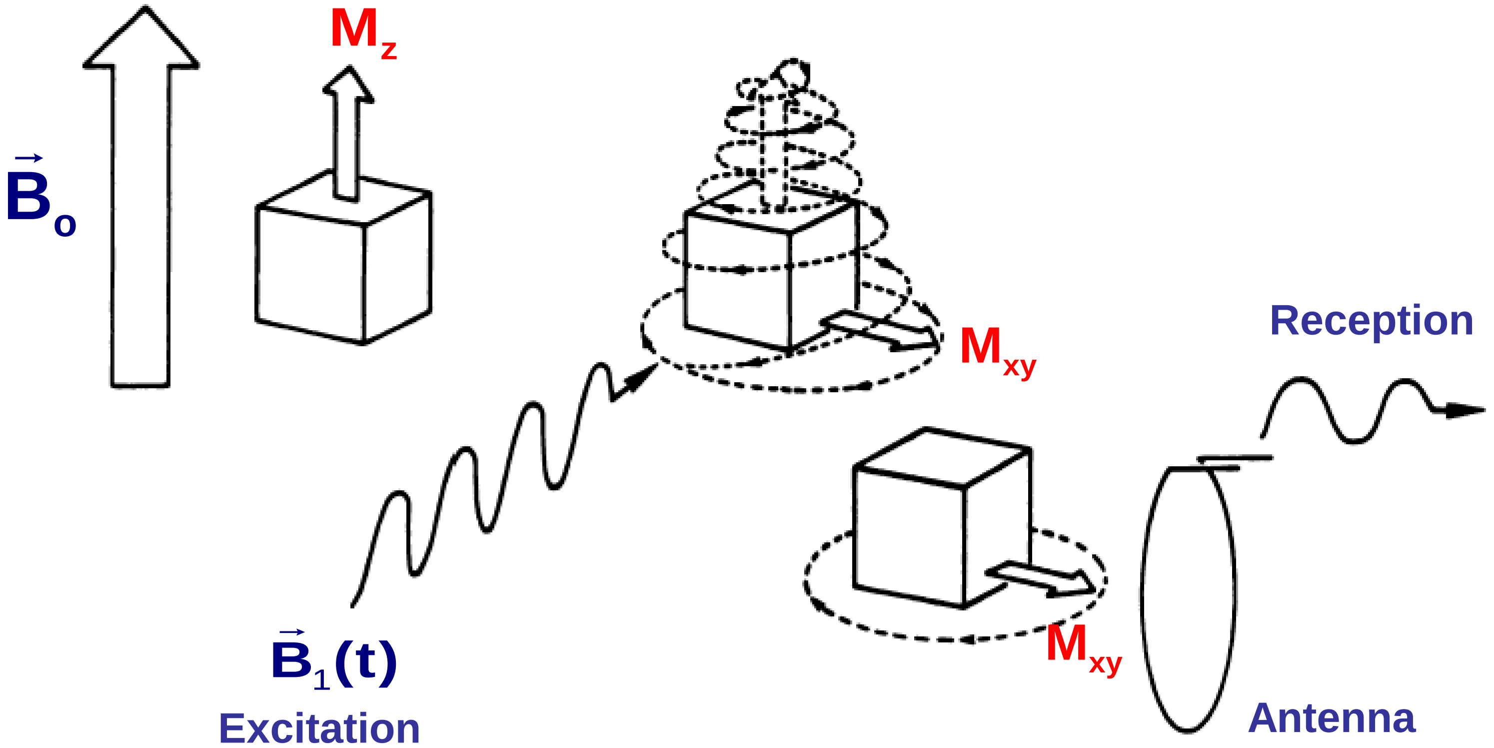
- **Magnetization returns to its equilibrium position → energy relaxation**



The amplitude of M_{xy} decreases as a function of time

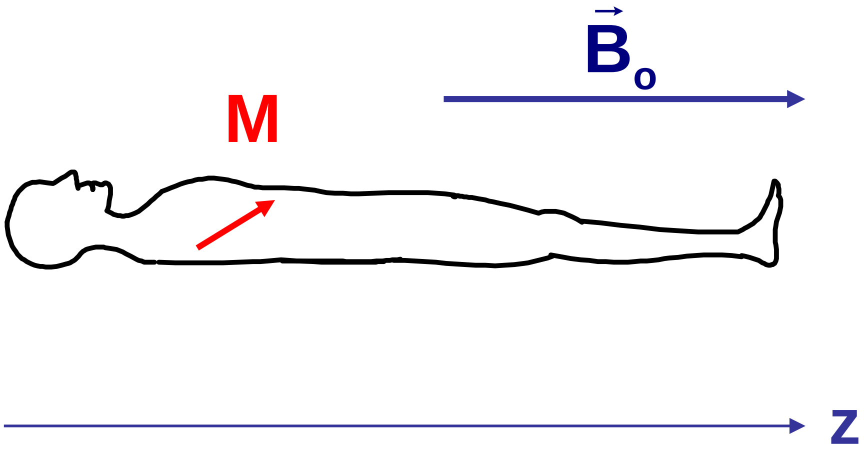
The amplitude of M_z increases as a function of time

Summary

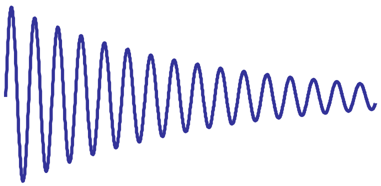


Basic contrast in MRI

- The speed of M_z growth along B_o depends on tissue
- It is characterized by the longitudinal relaxation (T1)
- Energy is transferred to the surrounding (spin-lattice)

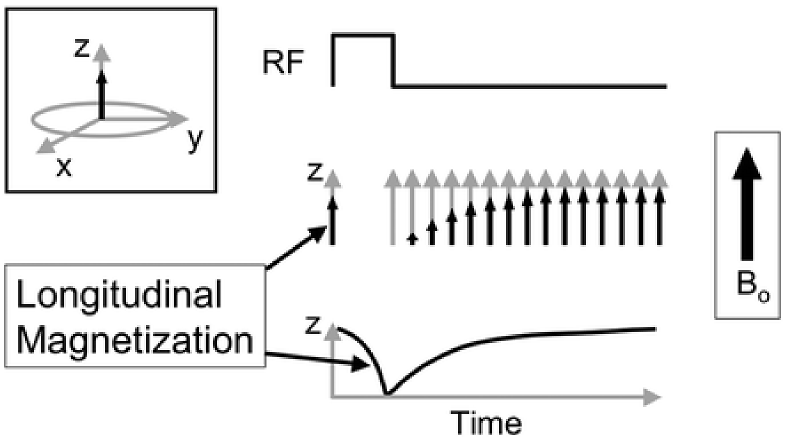


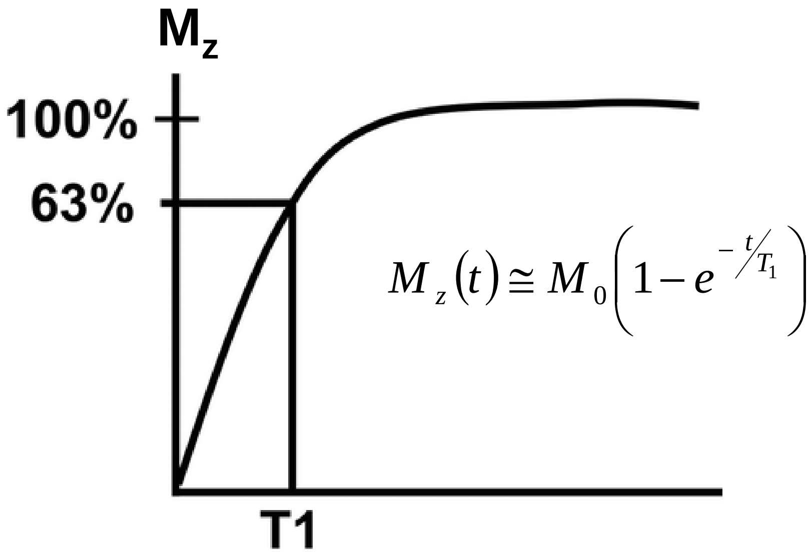
Recorded signal : $\mathbf{M_{xy}}$
(Demodulated)

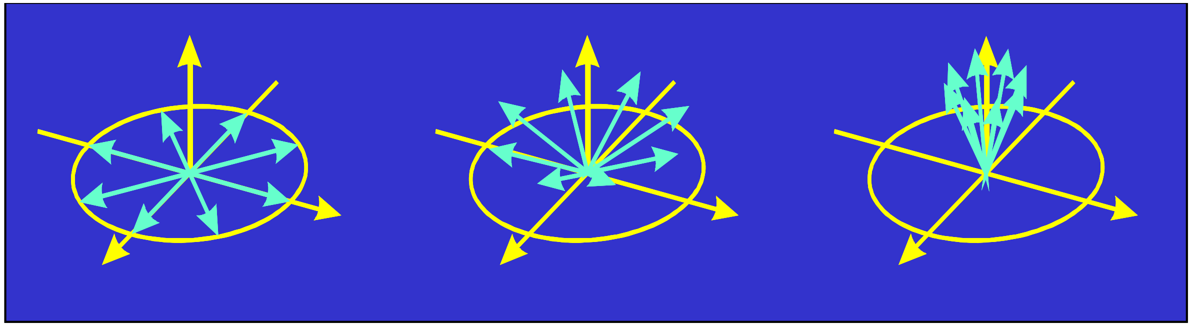


Longitudinal relaxation (T1)

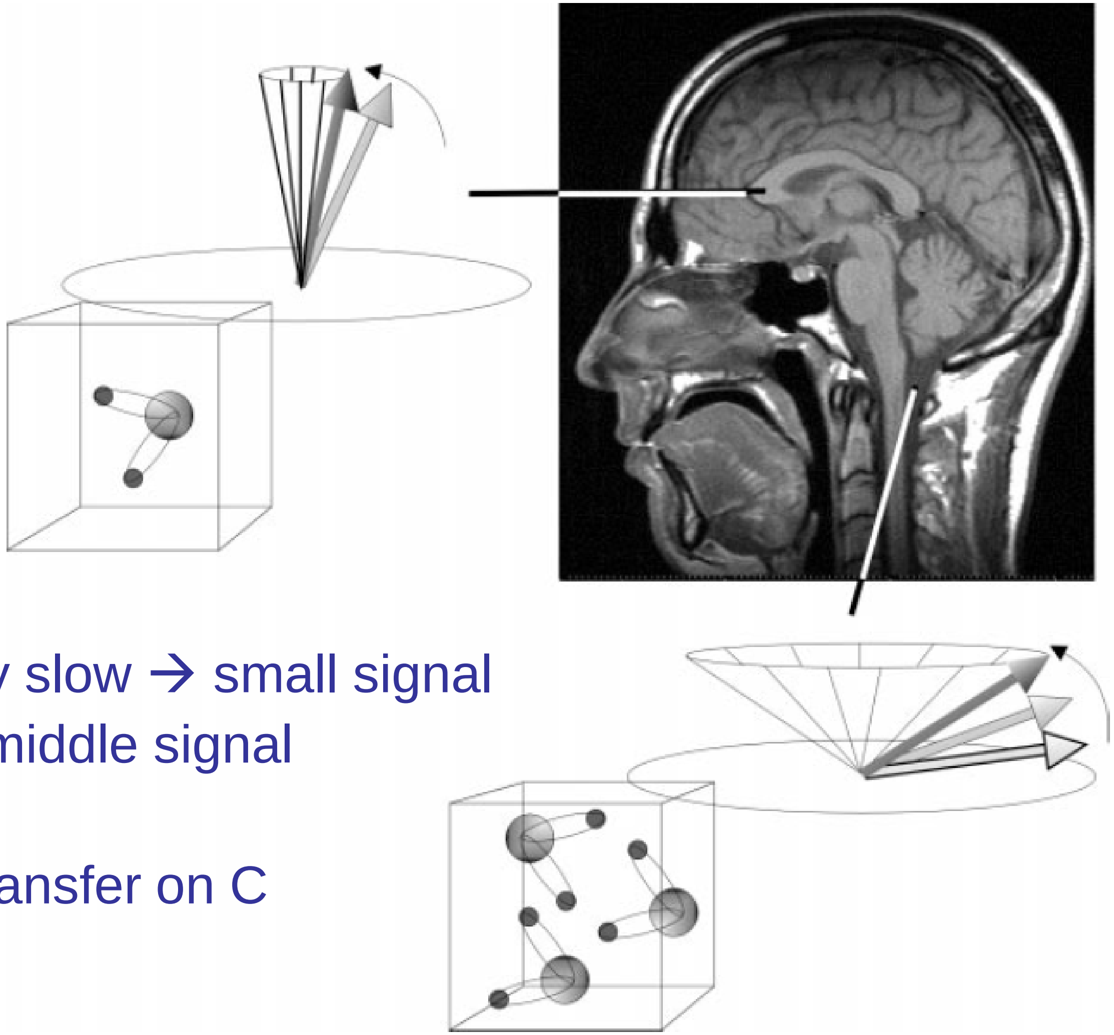
- Growth of M_z : magnetization that will be available to be tilted at the next excitation step → Origin of the MRI signal







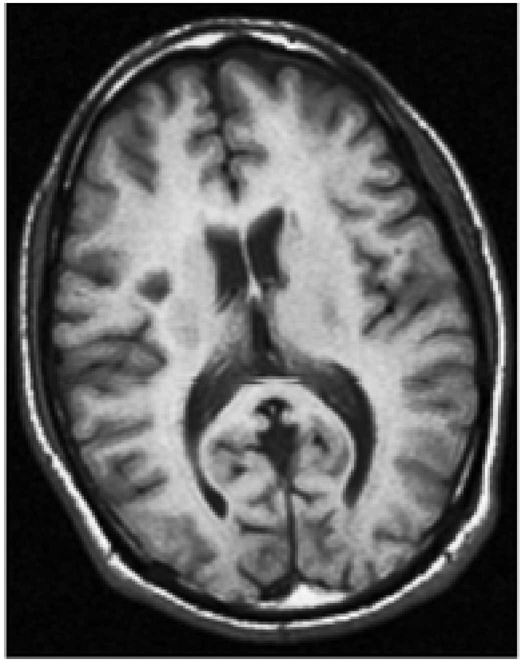
Longitudinal relaxation and T1 contrast

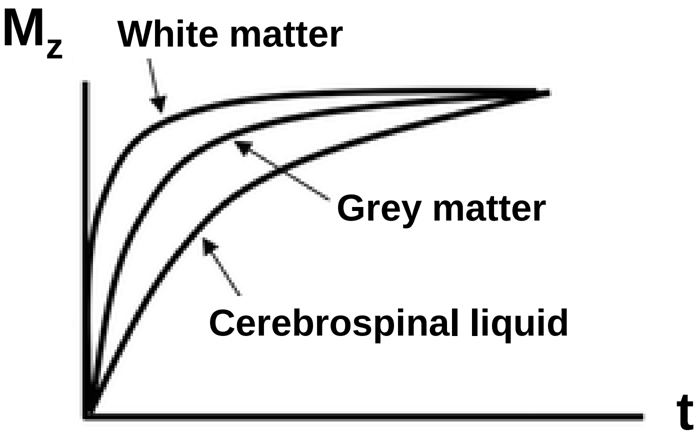


- Growth of M_z :
- Solids and liquids : very slow → small signal
- Soft tissue : middle → middle signal
- fat: fast → large signal
  - Magnetization transfer on C

Longitudinal relaxation and T1 contrast

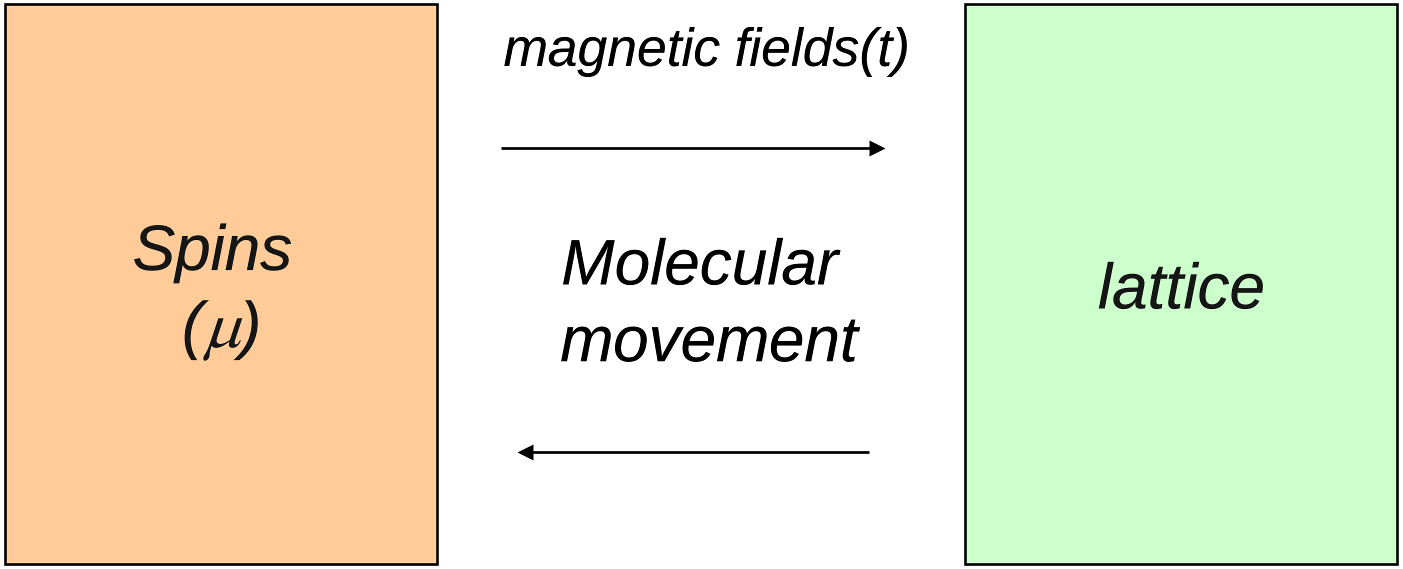
- T1 is tissue characteristic
- Imaging using T1 weighting





Longitudinal relaxation (T1) : magnetization growth along z (longitudinal axis)

Longitudinal relaxation and T1 contrast



Movements of μ induces magnetic field variations → stimulated relaxation

Longitudinal relaxation and T1 contrast

-Dipolar relaxation (other spins) $T1_d$

-Dipolar paramagnetic relaxation (Gd) $T1_p$

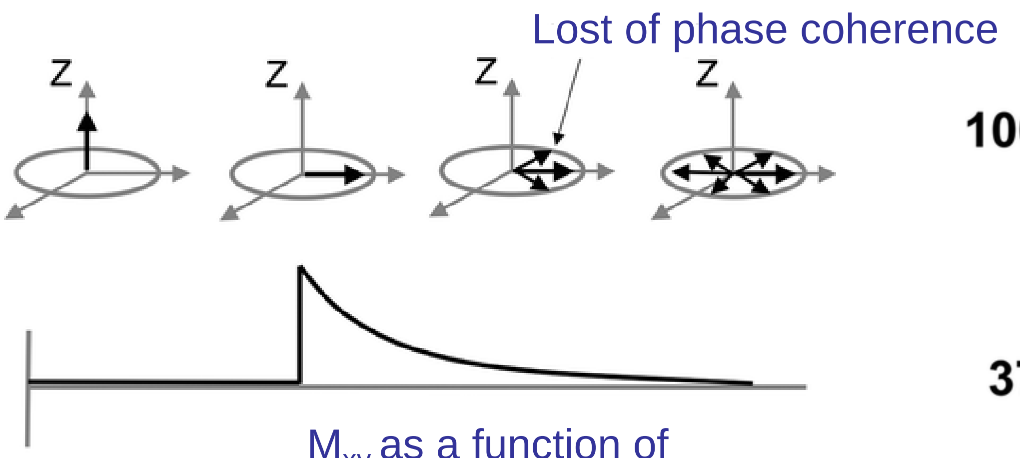
-Quadripolar relaxation T1 (spin > ½)

-…

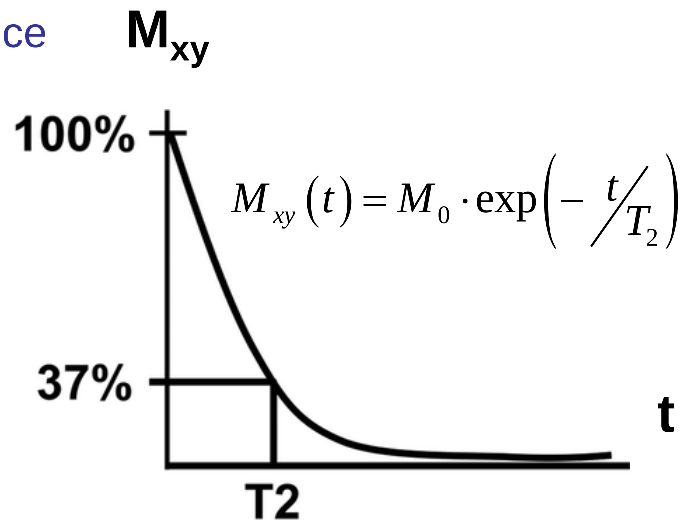
$$\frac{1}{T_1} = \sum \frac{1}{T_i}$$

Transverse « relaxation » (T2)

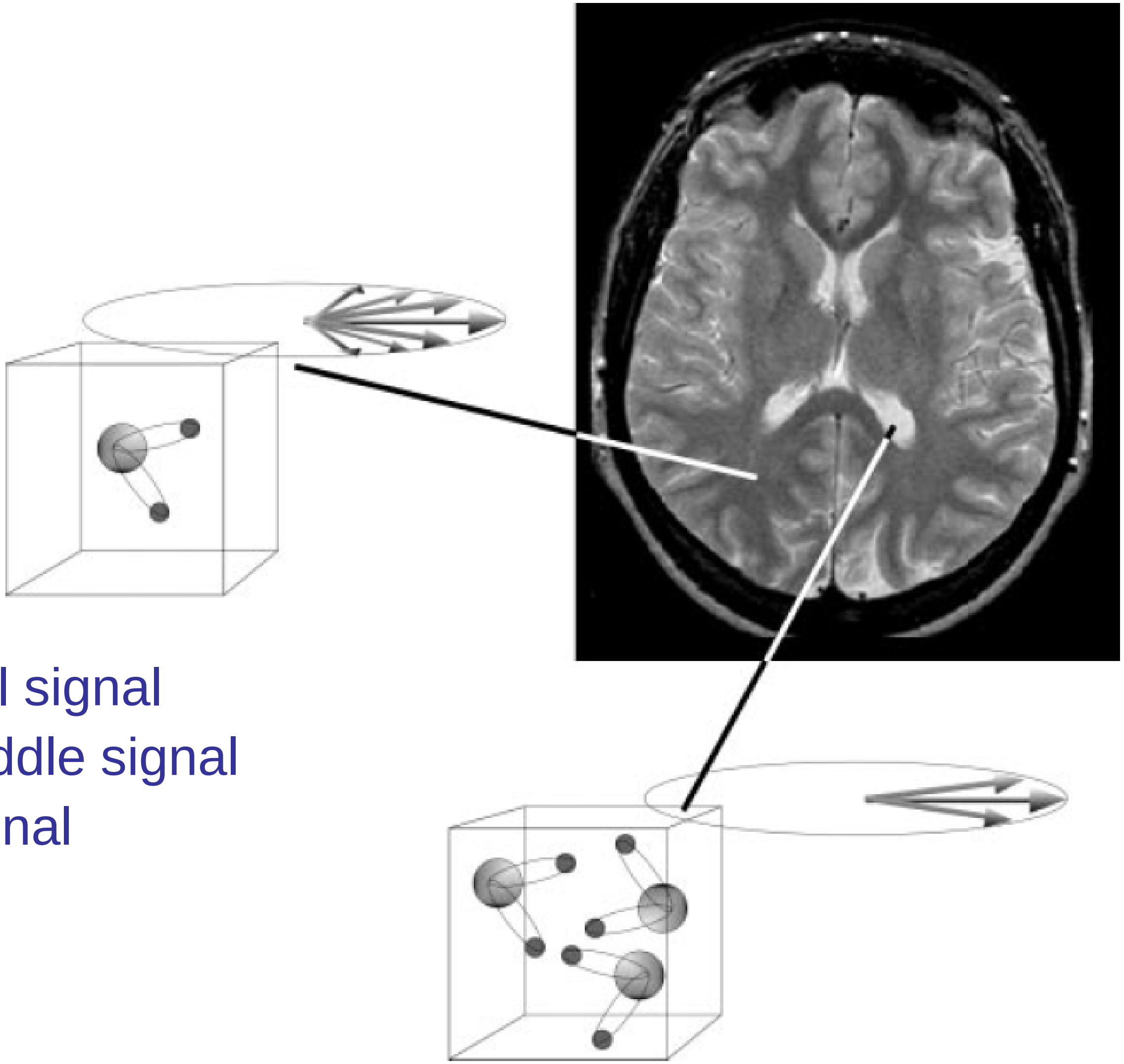
- Simultaneous with T1 but faster
- Lost of phase coherence between the individual spin
    - Decrease of M_{xy} (signal) (in addition of T1 effect)
    - No energy exchange with the lattice (spin-spin)
- Depends on the magnetic homogeneity of the tissue



M_{xy} as a function of time



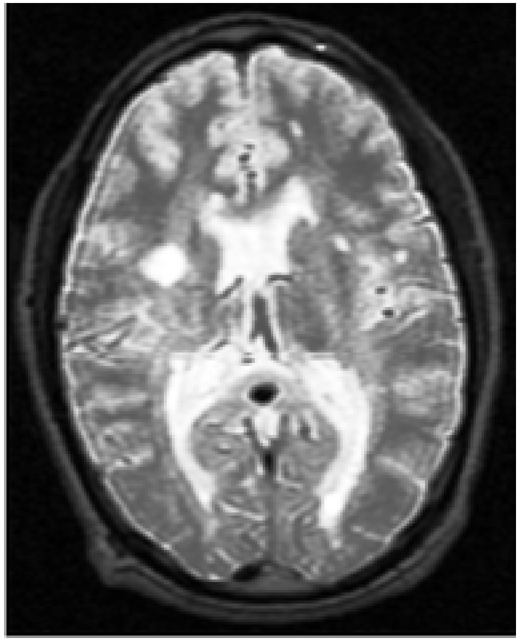
Transverse « relaxation » (T2)

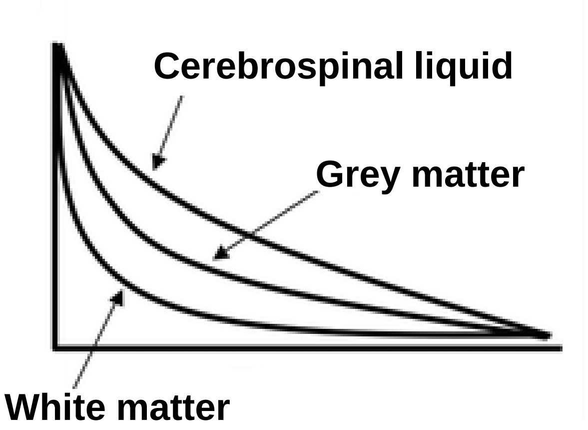


- Reduction of M_{xy} :
- Solids : very fast → small signal
- Soft tissue: middle → middle signal
- Liquids: slow → large signal

Transverse « relaxation » (T2)

- T2 is also a tissue characteristic
- Imaging using T2 weighting





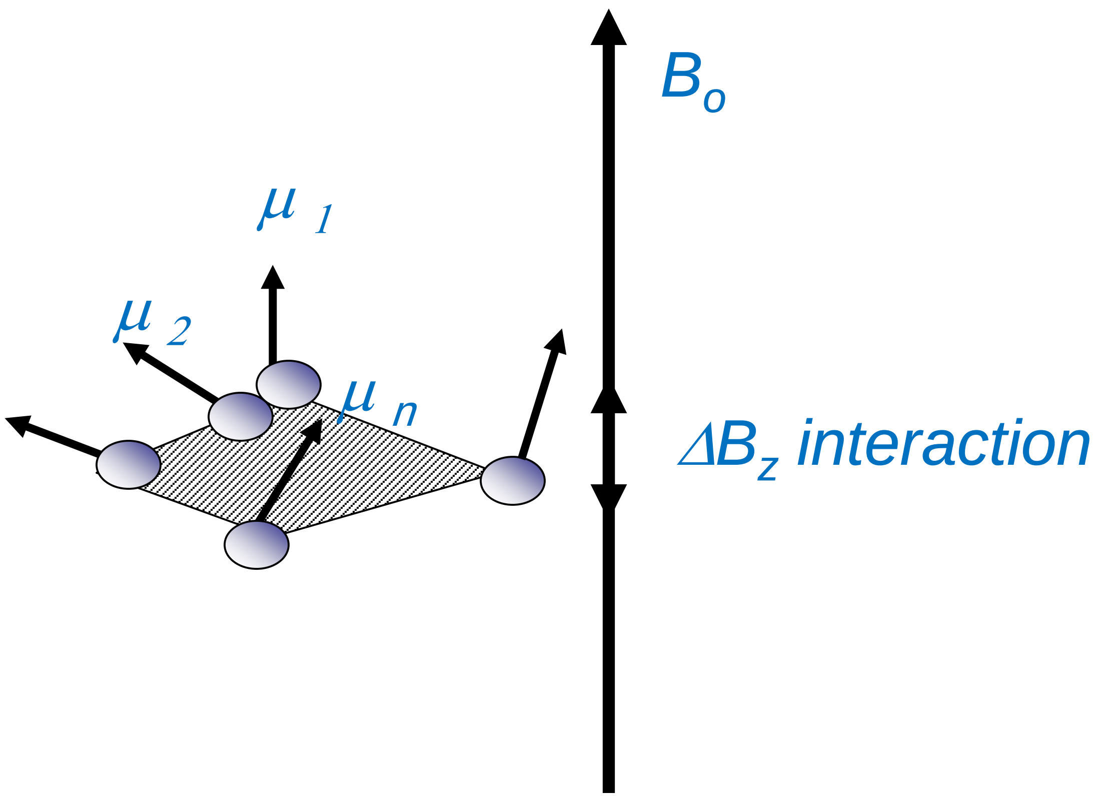
Order of magnitude

- **Indicative values (T1 in general >> T2)**

Tissue	T1 @1,5 T (msec)	T2 (msec)
Fat	260	80
Liver	500	40
Muscle	870	45
White matter	780	90
Grey matter	900	100
Cerebrospinal liq.	2’400	160

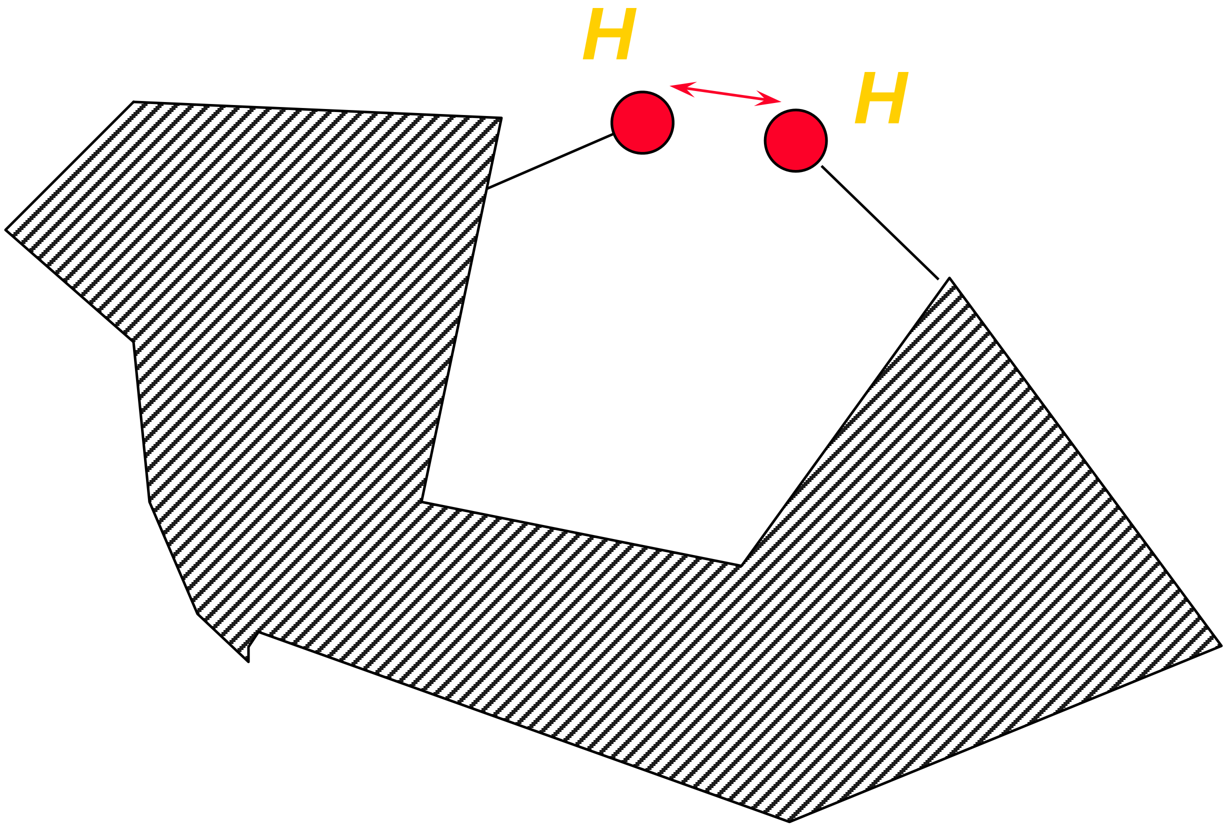
Transverse « relaxation » (T2)

Simultaneous with T1, but faster → lost of magnetization coherence
Mutual effects of μ



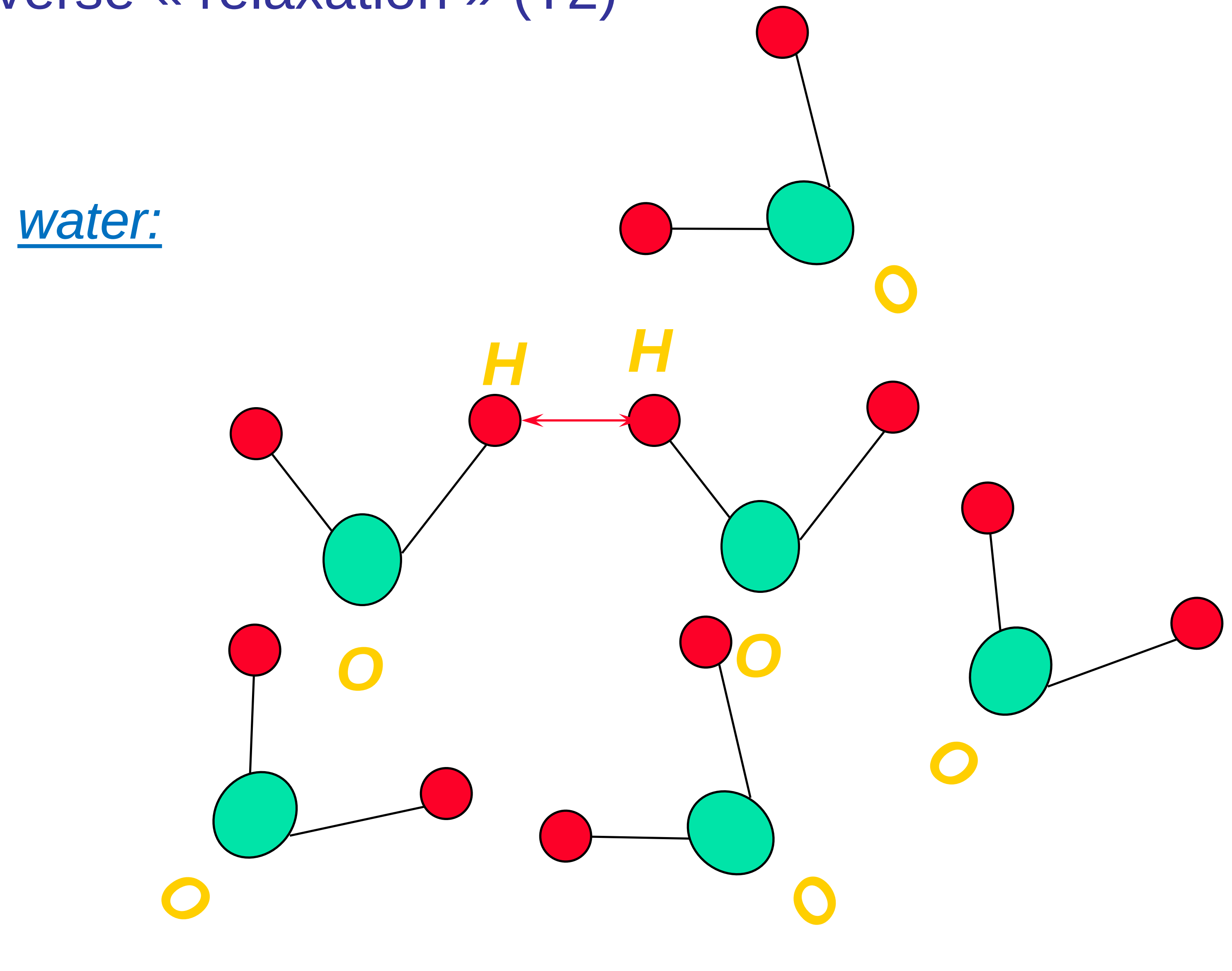
Transverse « relaxation » (T2)

Large molecules (protein)



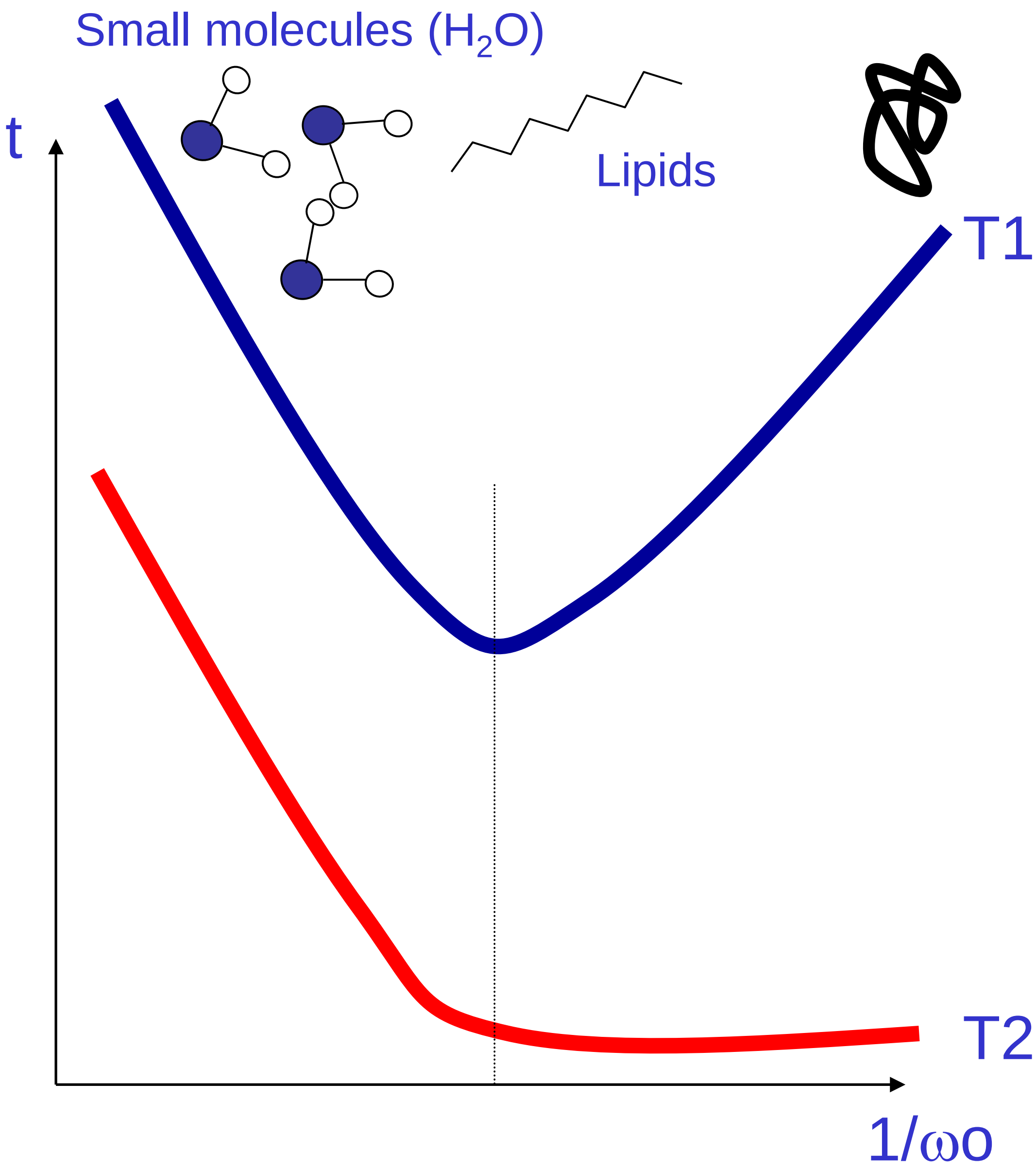
Slow movements → Short T2

Transverse « relaxation » (T2)



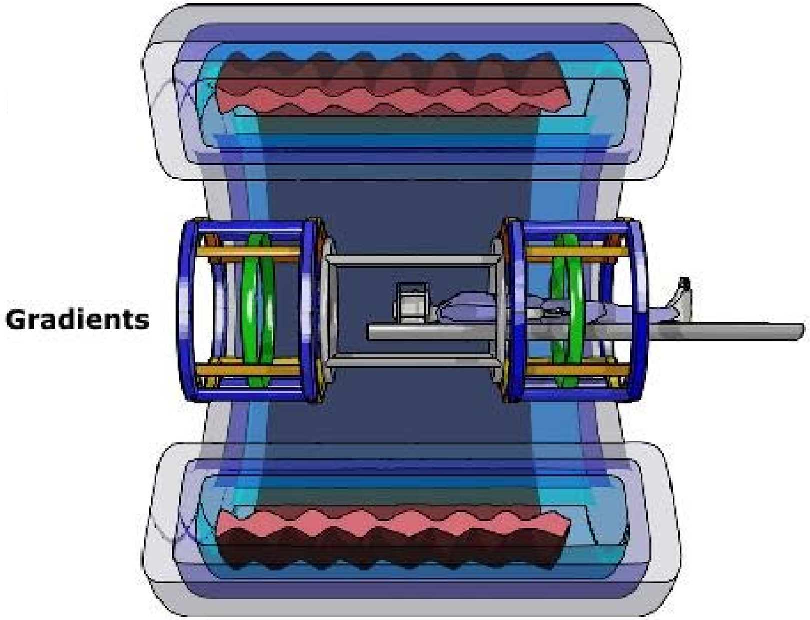
fast movements → averaging of field variations → longT2

T1 and T2 as a function of molecular dynamic

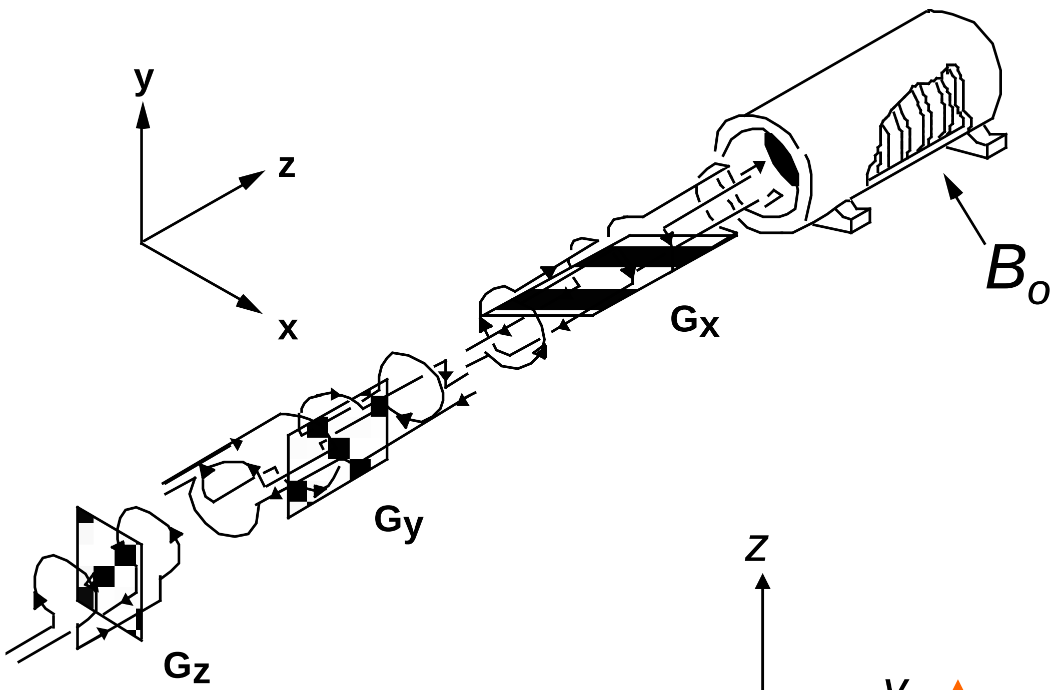


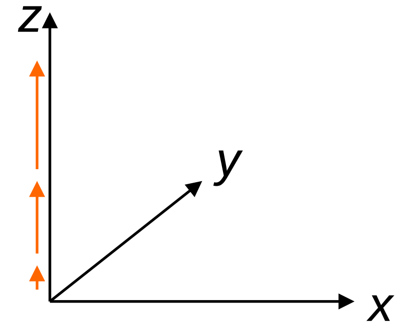
Schematic of an MRI system

- A large magnet (static magnetic field B_o (0.5 to 3 T in clinic)
- Three gradient coils (magnetic field B_o that varies with position x,y,z) (a few mmT/m)
- Excitation and reception antenna (magnetic field that varies with time)

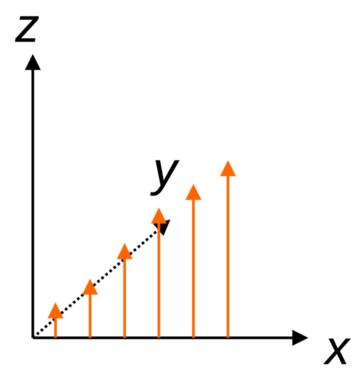


From NMR to MRI

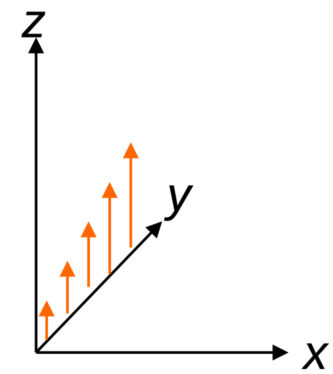




Z gradient

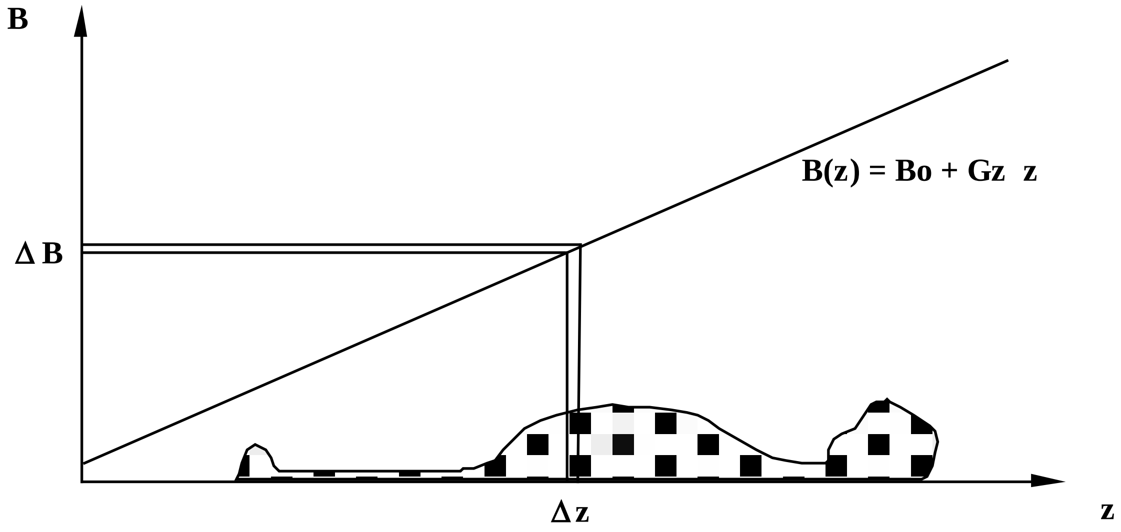


X gradient



Y gradient

Slice selection



G_z a few $mT.m^{-1}$

Express the slice thickness D_z as a function of G_z

Tomographic technique → how to choose a slice?

Slice position : frequency value

Frequency domain

$$\Delta\omega = \gamma \,.\, G_z \,.\, \Delta z$$

Bandwidth of the rf : $\Delta\nu = \Delta\omega/2\pi$

↓

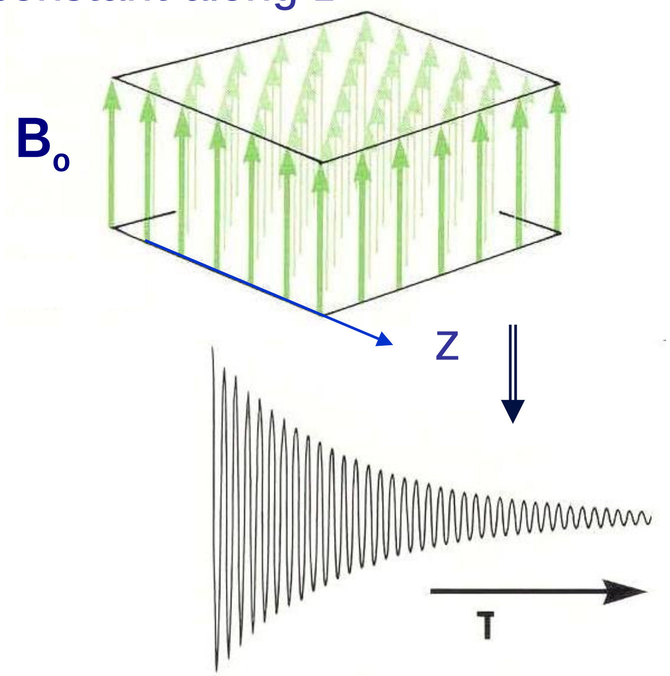
Slice thickness :

$$\Delta z = 2\pi\, \Delta\nu \,/\, (\gamma \,.\, G_z)$$

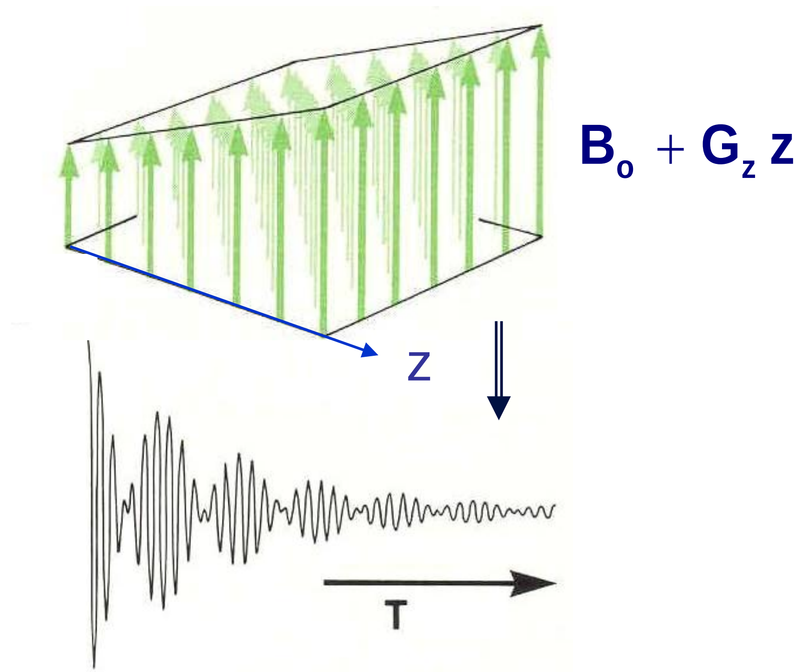
Tomographic technique → how to choose a slice?

- During the excitation apply a magnetic gradient field ⊥ to the slice plane
- Selective rf excitation to only tilt spin within the selected slice

The amplitude of B_o is constant along z



Only one Larmor frequency value

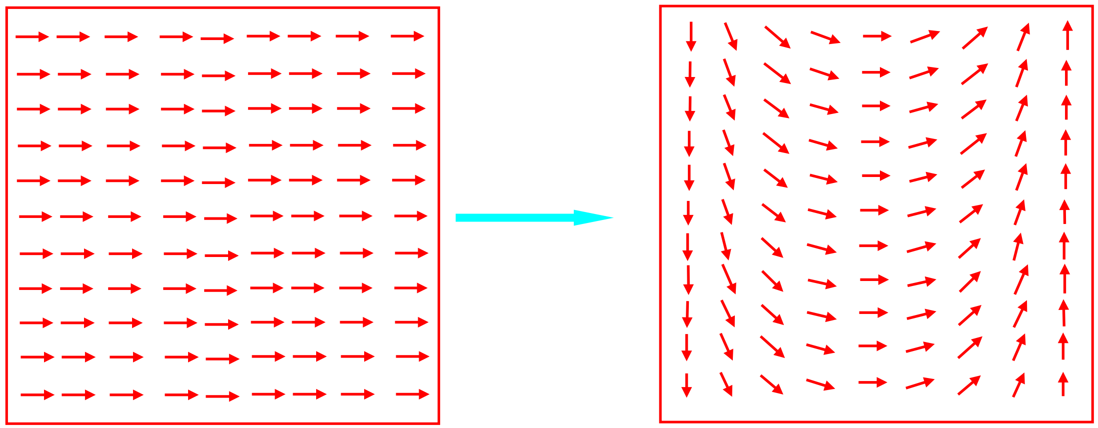


Several frequency Larmor frequencies

Encoding of the information

Phase space (k-space)

9×9 pixels

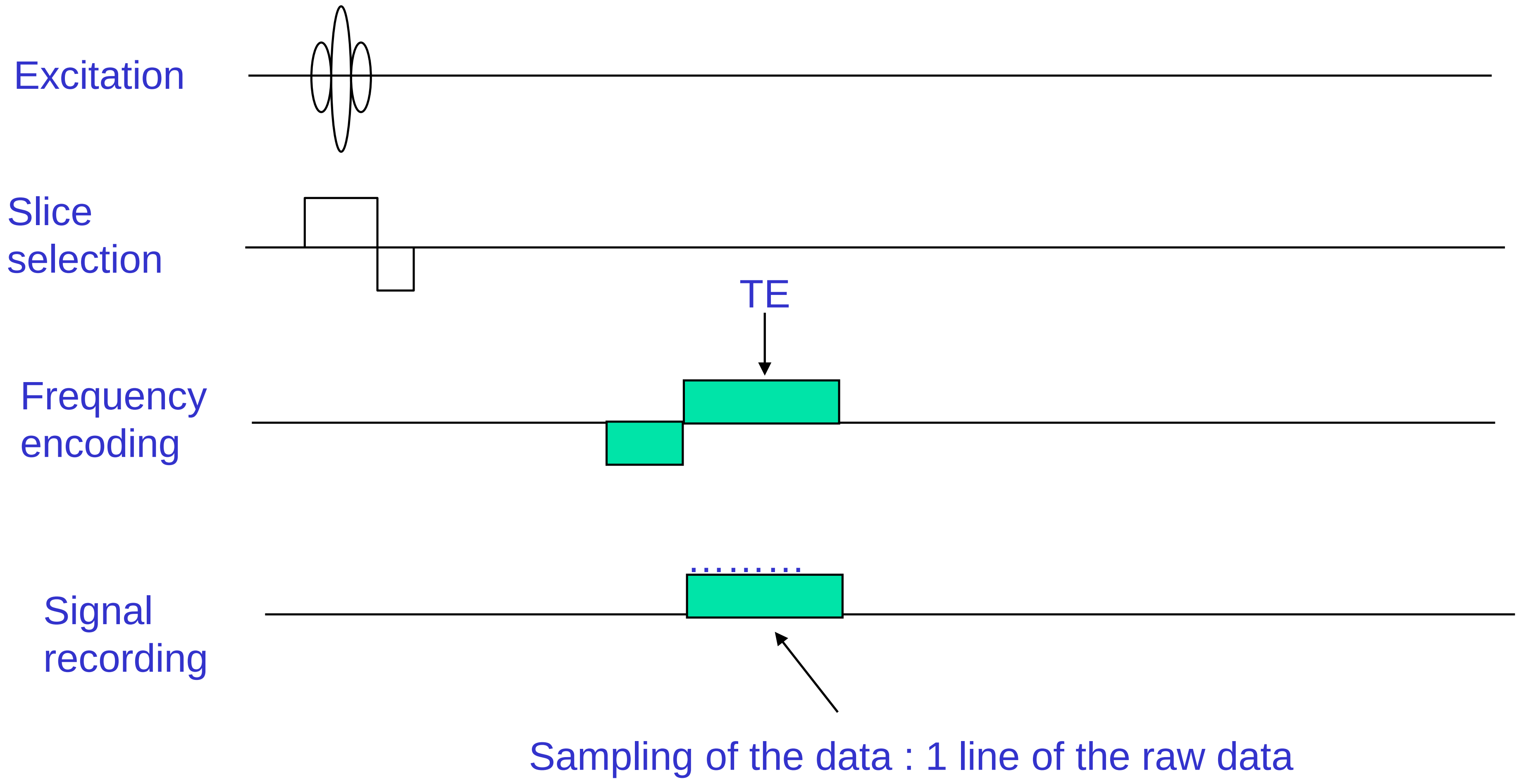


Before encoding

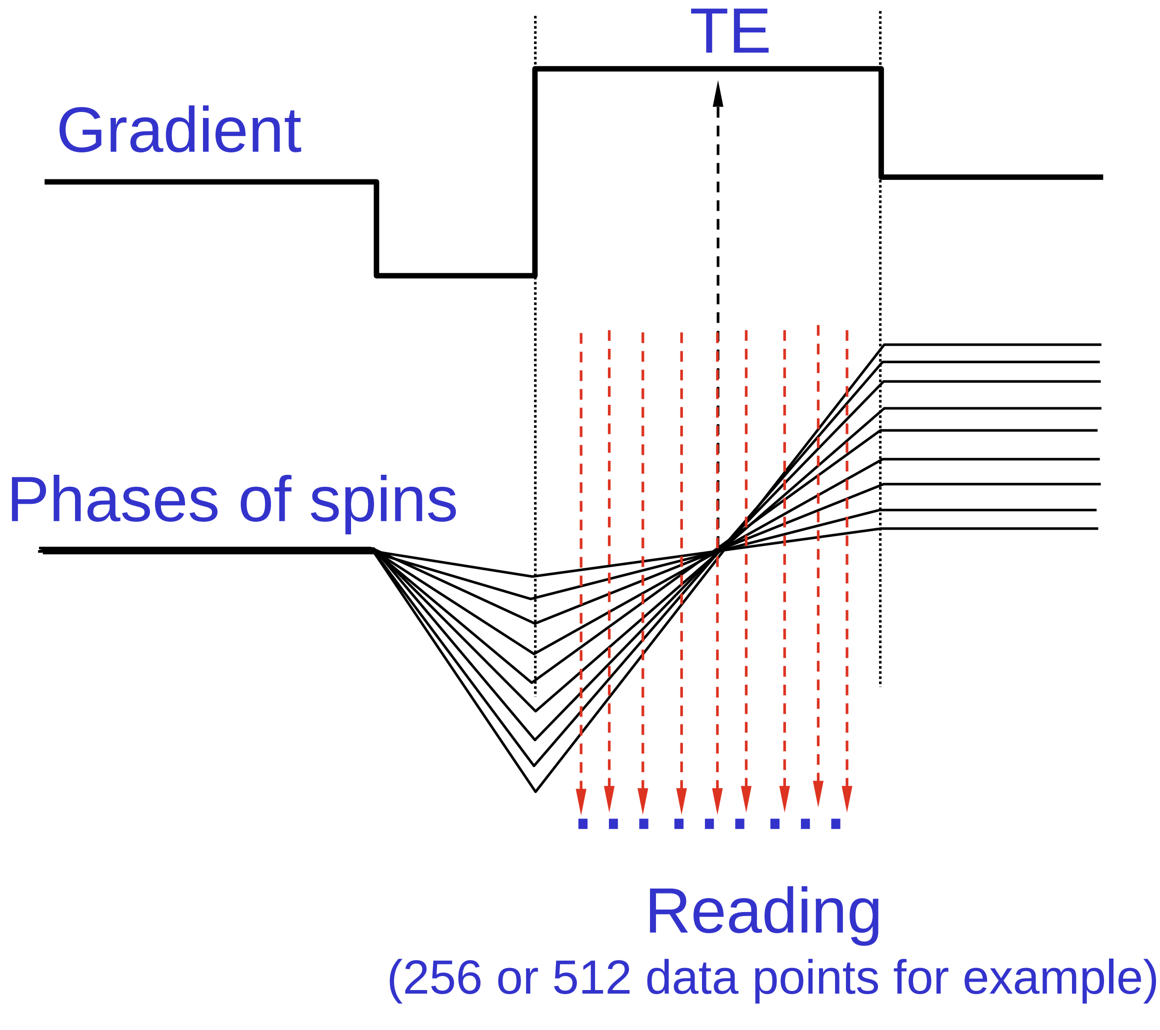
After encoding
(use of two other gradients than slice selection)

Goal to code each pixel position in terms of frequency and phase of the spin

Frequency encoding : during data acquisition

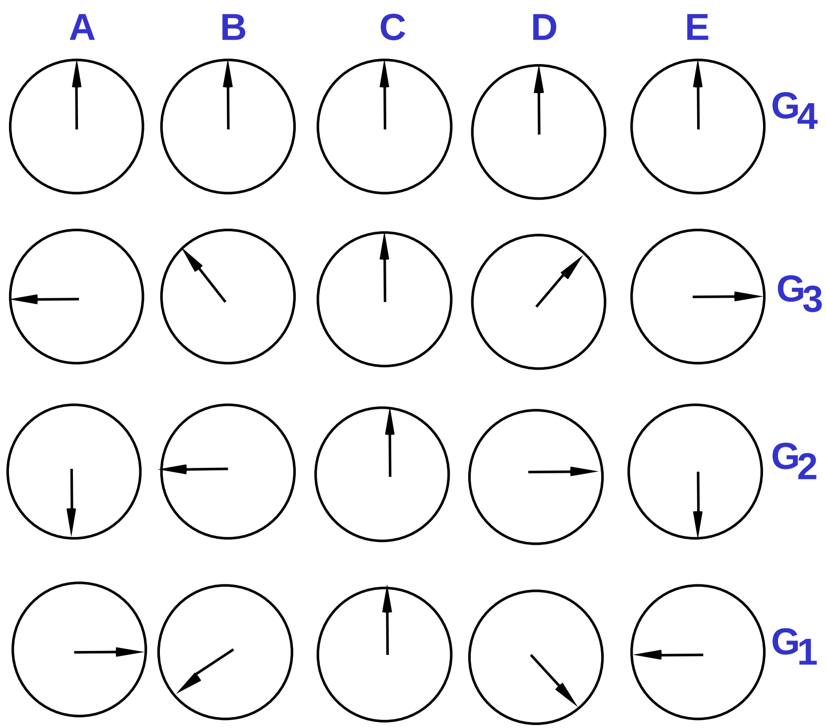


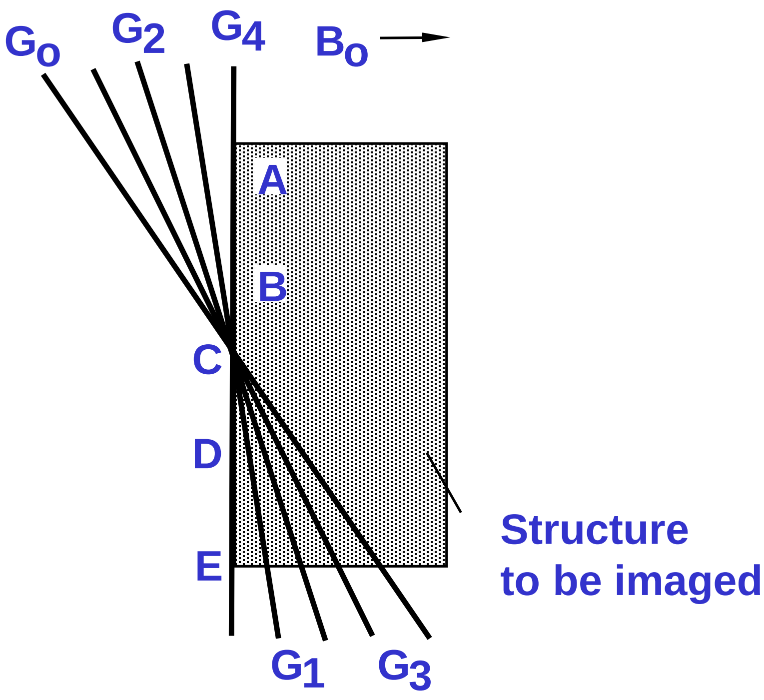
Frequency encoding : during data acquisition



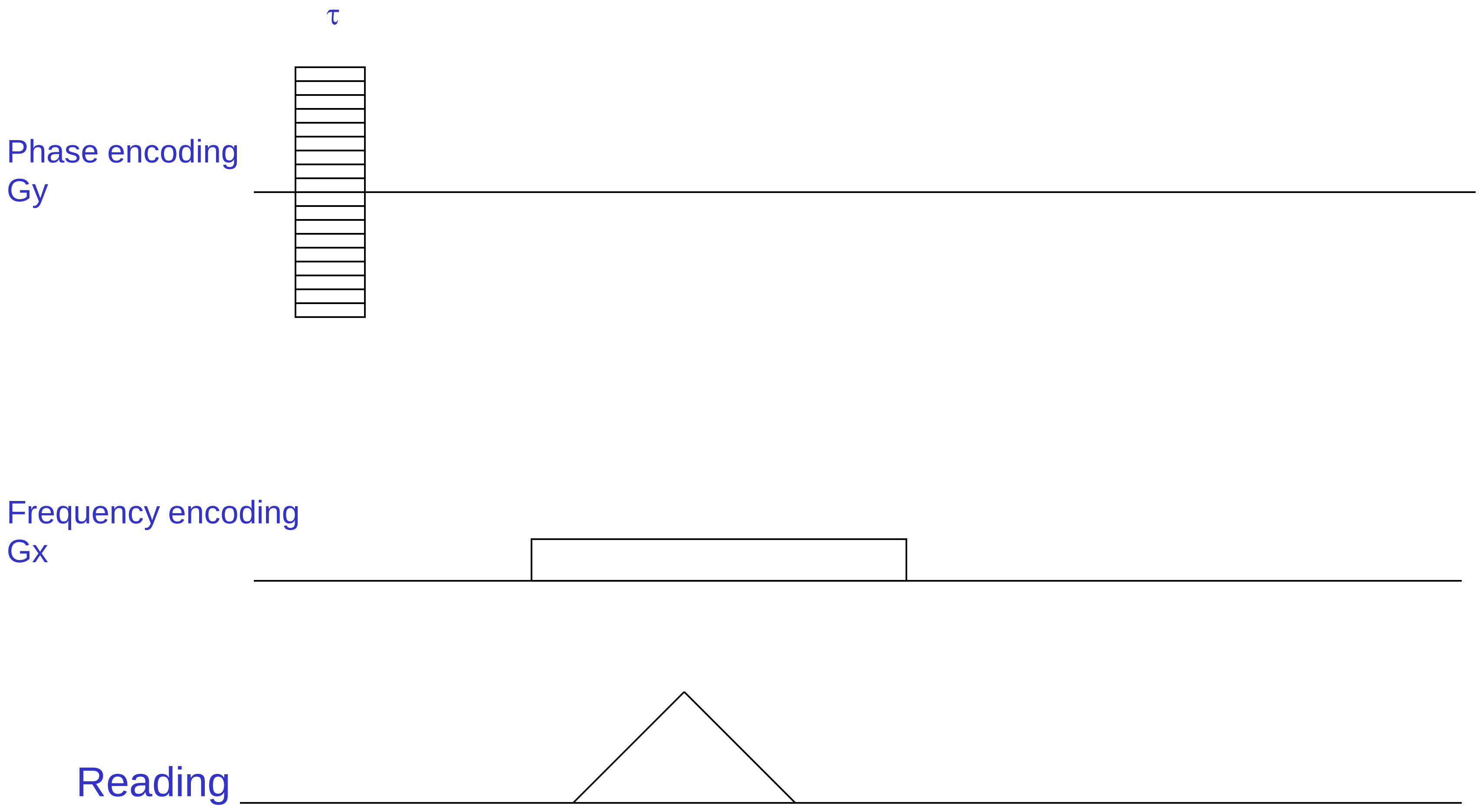
Phase encoding : before data acquisition

Each spin a tagged with a phase as a function of its position



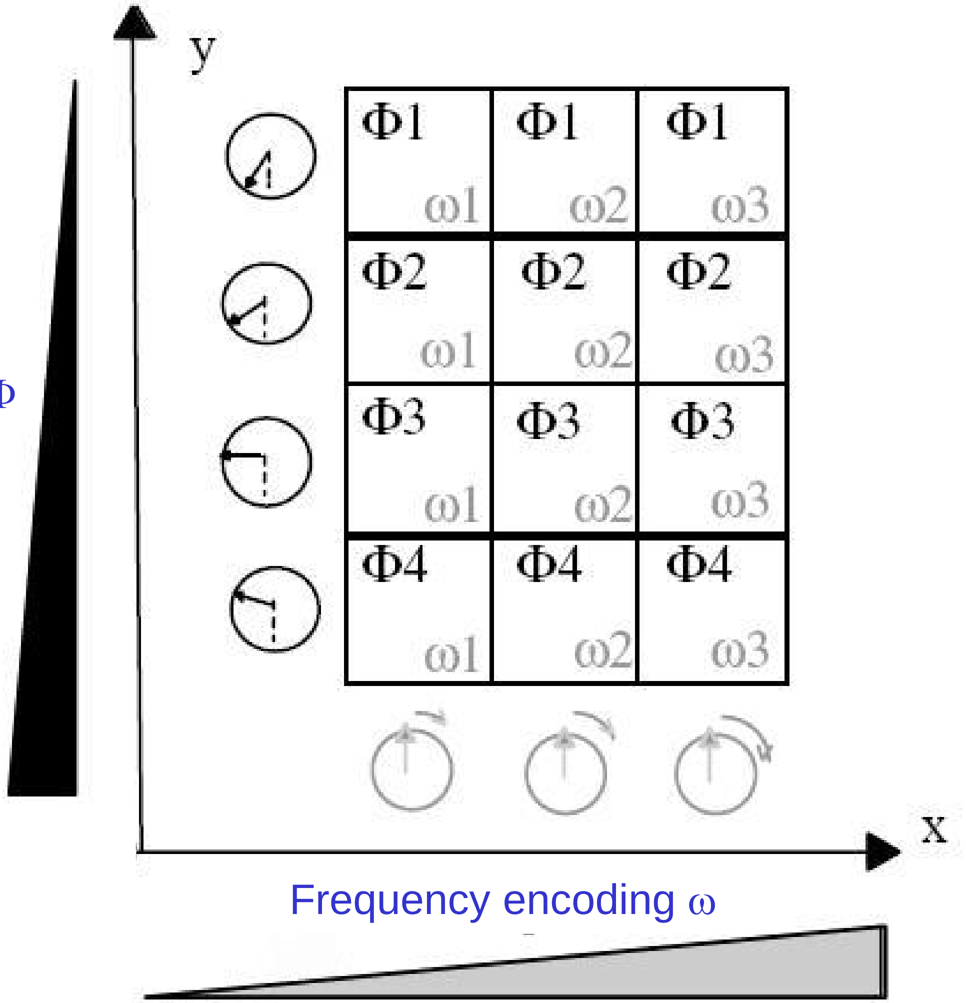


Part of the sequence to encode the image



Result of image encoding

Phase encoding Φ



Frequency encoding ω

Mathematics of the spin-warp

$$S(t, G_y) = \iint M(x, y)\, e^{2\pi i \gamma G_x x t} \cdot e^{2\pi i \gamma G_y y \tau}\, dx\, dy$$

$\mathrm{t\ and\ G_y\ produce\ phase\ differences: only\ one\ type\ of\ variable}$

$$\mathrm{k_x} = \gamma G_x t \quad \rightarrow t = n\Delta t = \frac{n\Delta \mathrm{k_x}}{\gamma G_x}$$

$$\mathrm{k_y} = \gamma G_y \tau \quad \rightarrow G_y = \frac{m\Delta \mathrm{k_y}}{\gamma \tau}$$

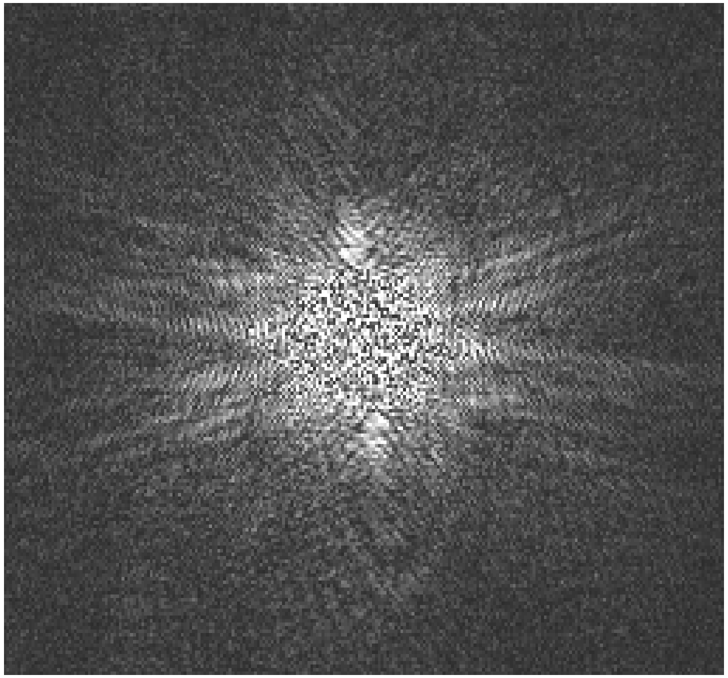
$$S(n\Delta \mathrm{k_x}, m\Delta \mathrm{k_y}) = \sum\sum M(x, y)\, e^{2\pi i (xn\Delta \mathrm{k_x} + ym\Delta \mathrm{k_y})}\, dx\, dy$$

$$\Delta \mathrm{k_x} = \frac{1}{N_x \Delta x} = \frac{1}{X} \qquad \Delta \mathrm{k_y} = \frac{1}{N_y \Delta y} = \frac{1}{Y}$$

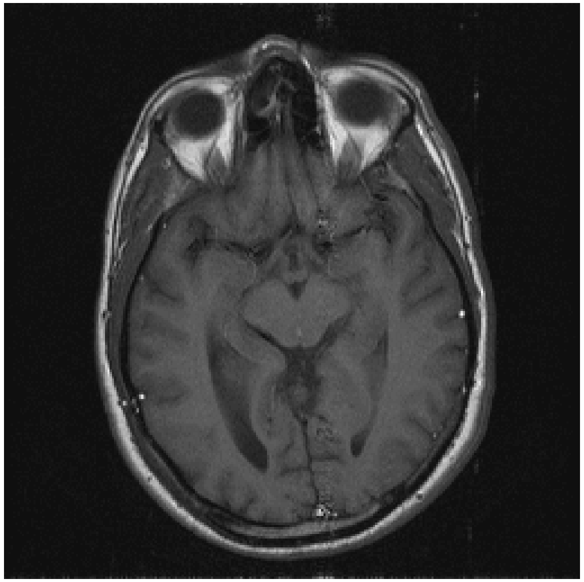
This method produce a elegant way of coding the data since it directly produces the Fourier transform of the image

$$M(x, y) = \sum\sum S(n\Delta \mathrm{k_x}, m\Delta \mathrm{k_y})\, e^{-2\pi i (xn\Delta \mathrm{k_x} + ym\Delta \mathrm{k_y})}\, \Delta \mathrm{k_x} \Delta \mathrm{k_y}$$

Summary

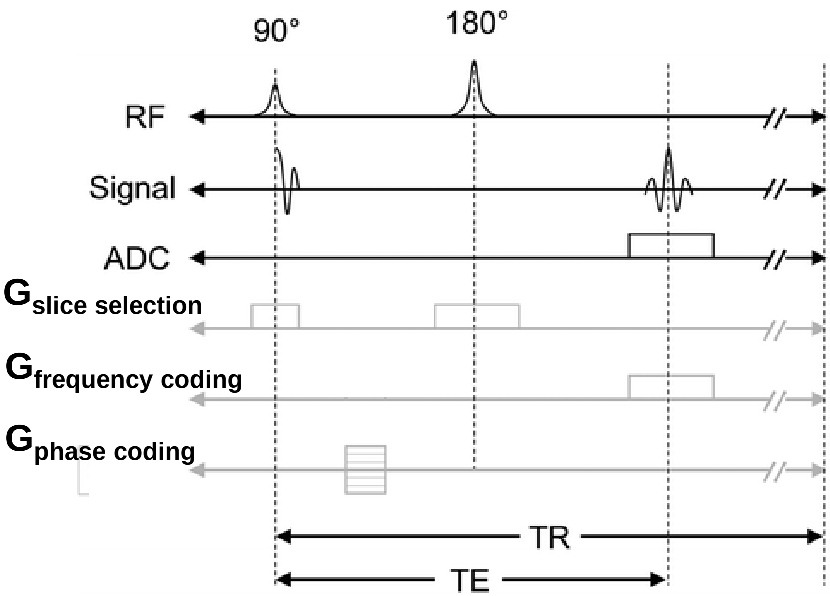


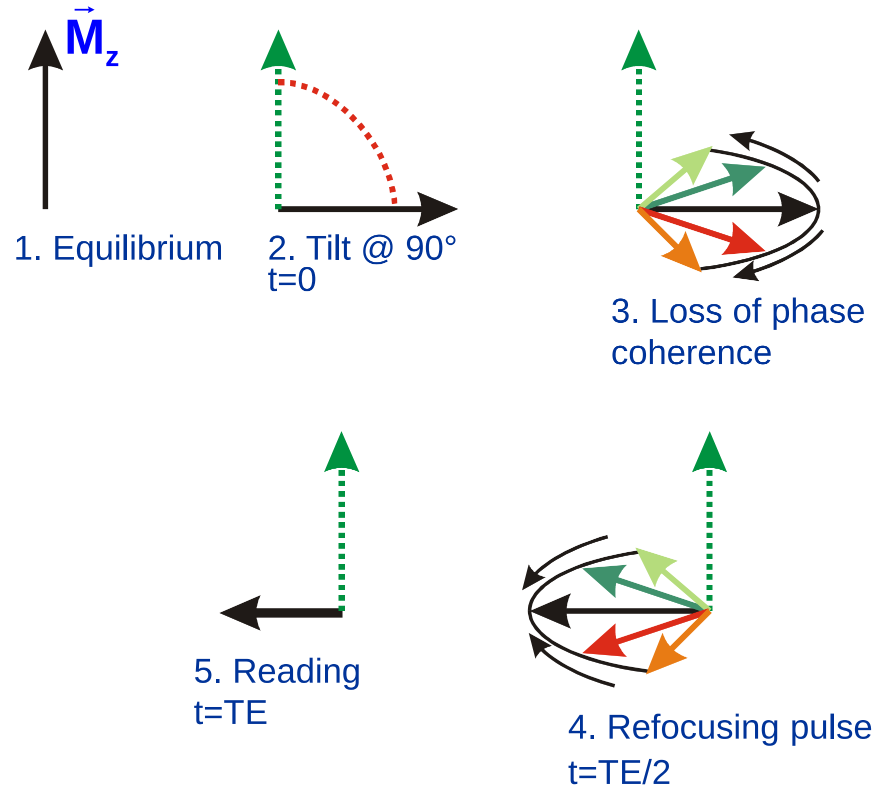
Fourier transform



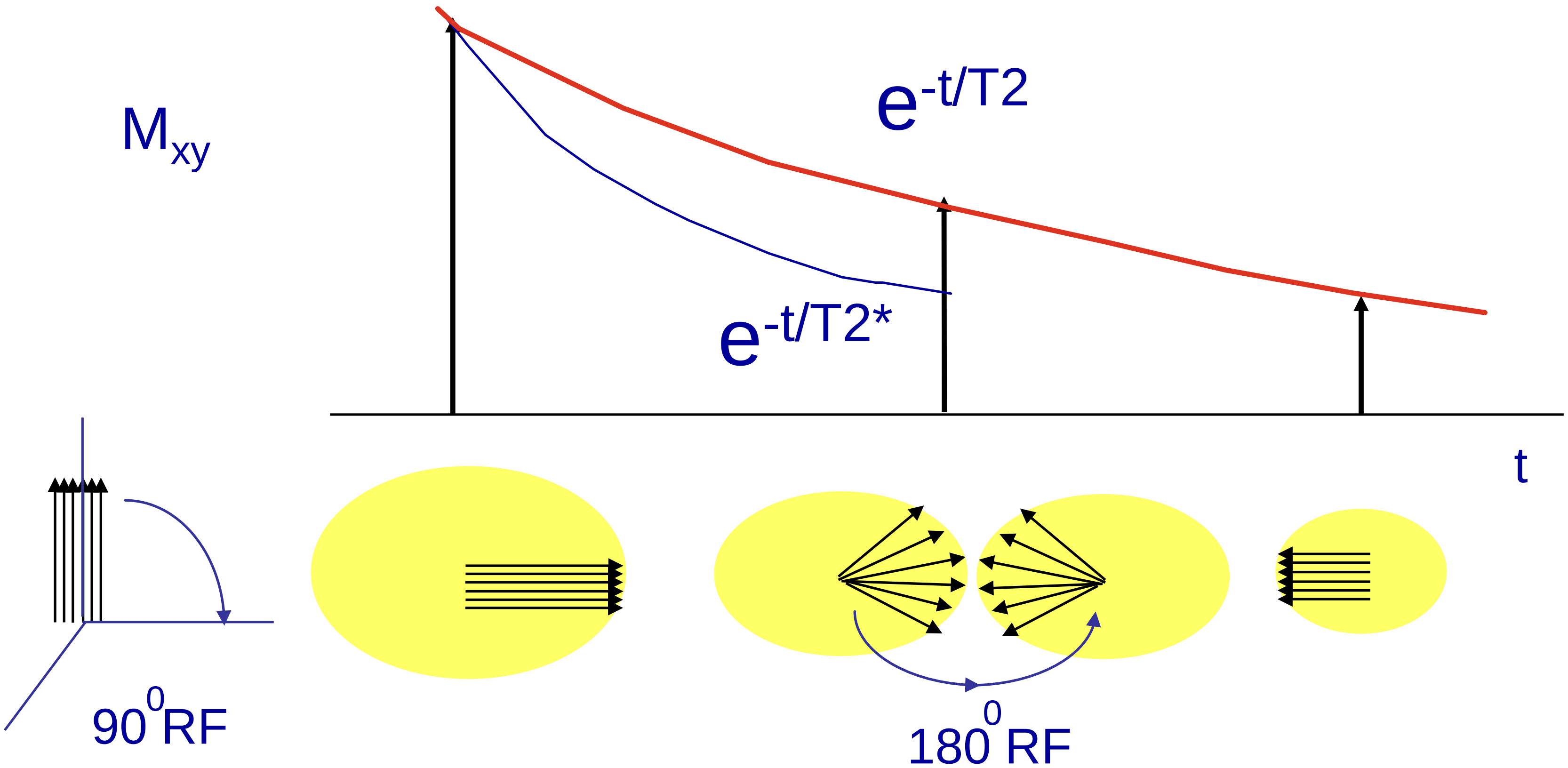
$$M(x,y) = \sum\sum S(n\,\Delta \mathrm{k}_\mathrm{x}, m\,\Delta \mathrm{k}_\mathrm{y})\, e^{-2\pi i (xn\,\Delta \mathrm{k}_\mathrm{x} + ym\,\Delta \mathrm{k}_\mathrm{y})}\, \Delta \mathrm{k}_\mathrm{x} \Delta \mathrm{k}_\mathrm{y}$$

Basic sequence : spin echo





Basic sequence : spin echo

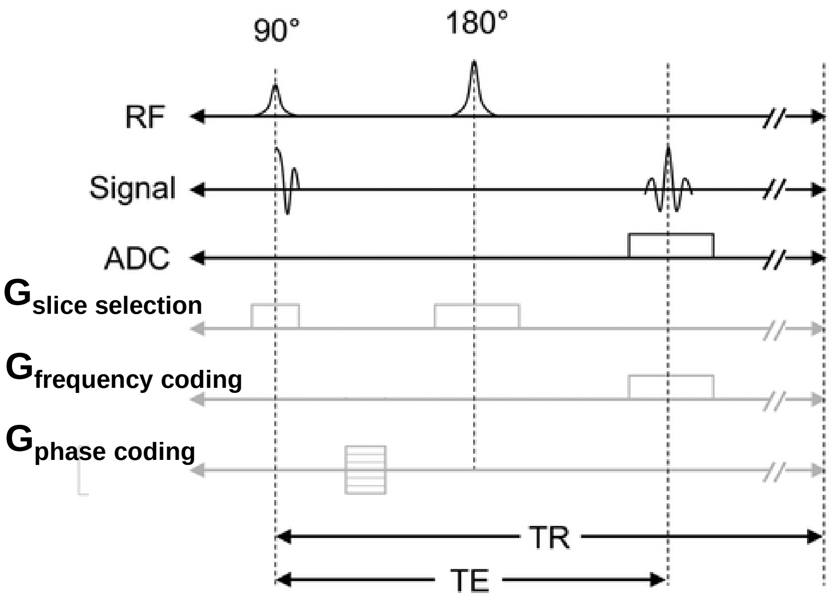


MRI sequence

- Sequence of rf and magnetic gradient as a function of time to build the image aiming to enhance a particular contrast:
  - T1 contrast (image of the anatomy)
  - T2 contrast (“ + « pathology »)
  - What would be the effect of an oedema in T2 ?

MRI sequence

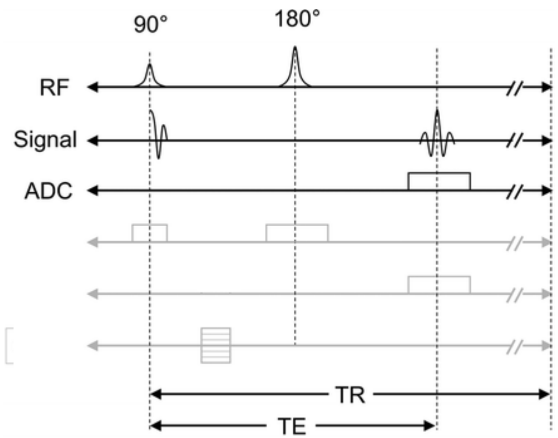
Control over the image contrast

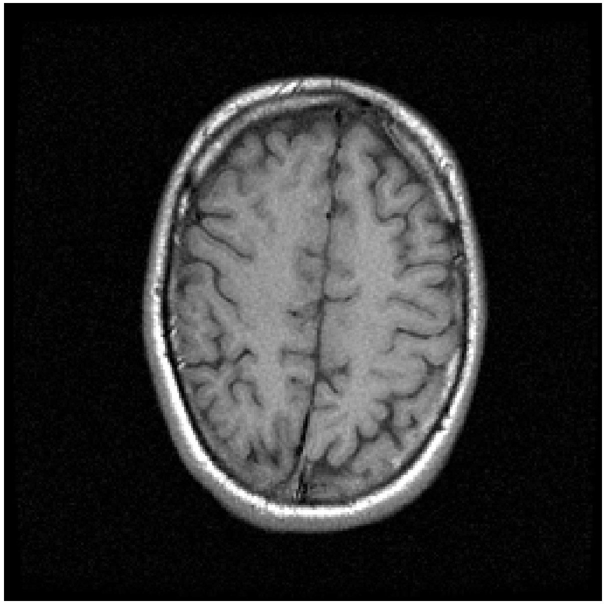


		TR	
		Short	Long
TE	Short	T1	Protons density
	Long	T1, T2	T2

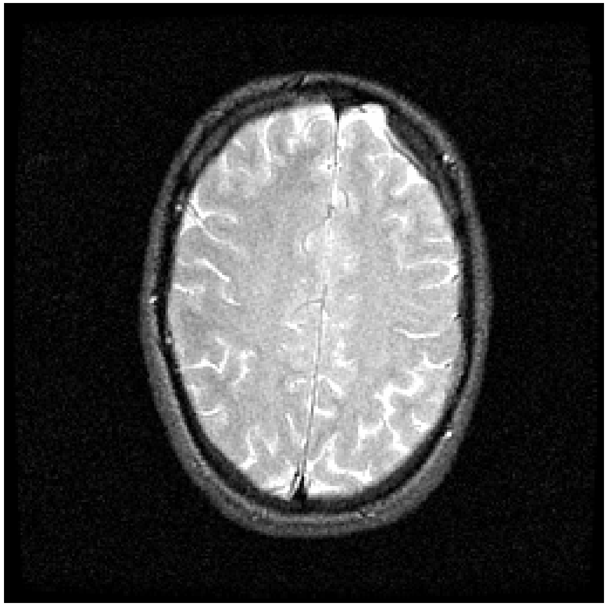
$$S(TR, TE) \propto \rho (1 - e^{-TR/T_1}) e^{-TE/T_2}$$

Contrast variation

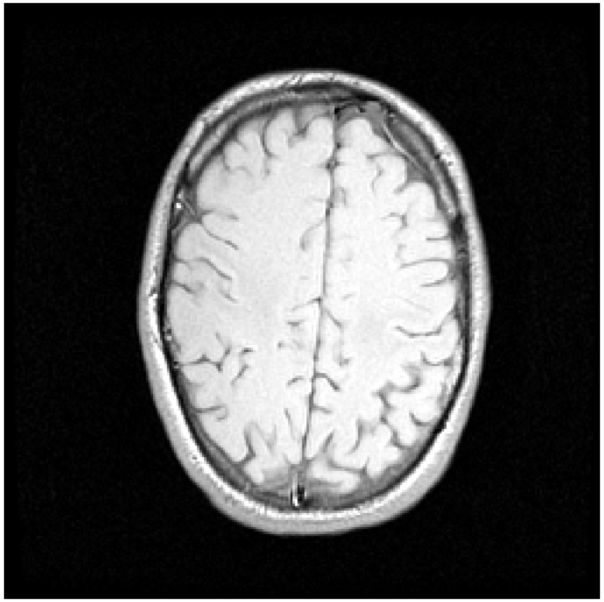




T1 Contrast
T_E = 14 ms
T_R = 400 ms

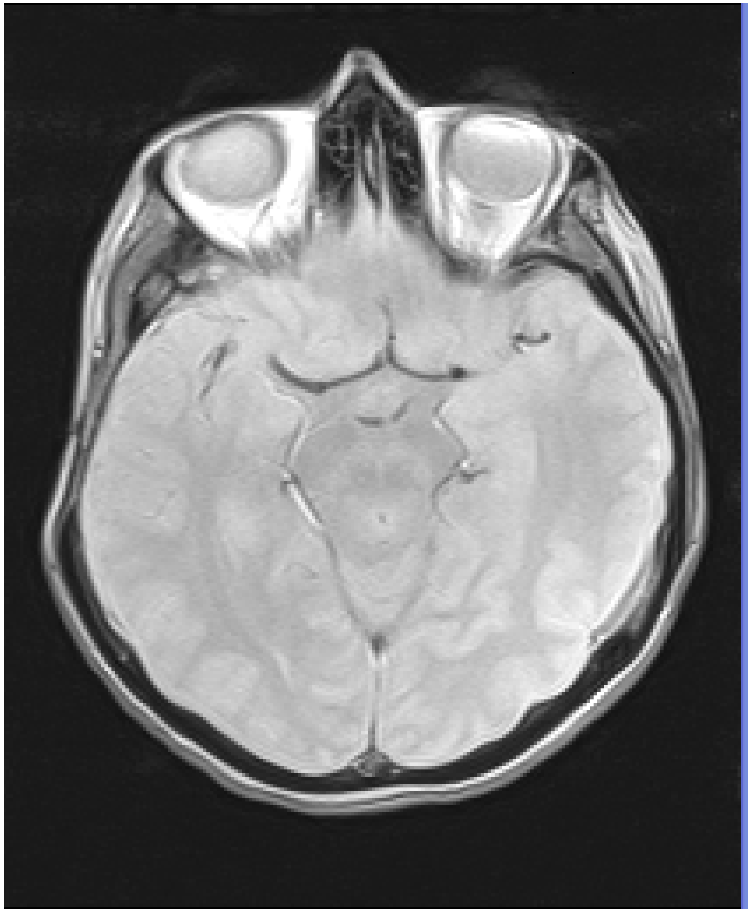


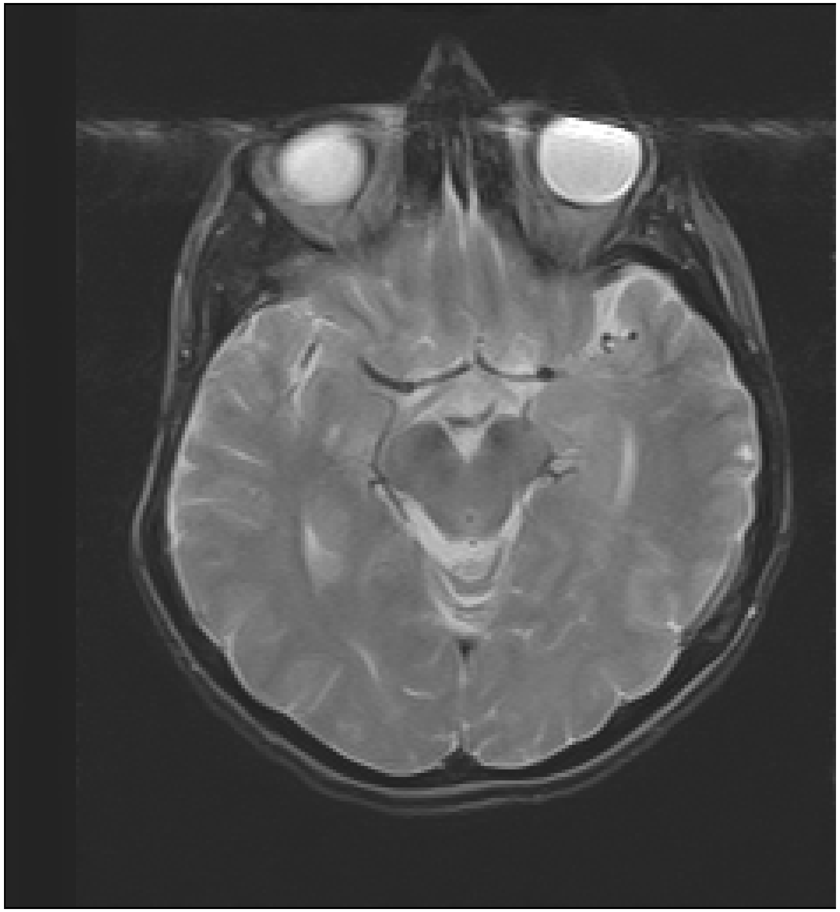
T2 Contrast
T_E = 100 ms
T_R = 1500 ms



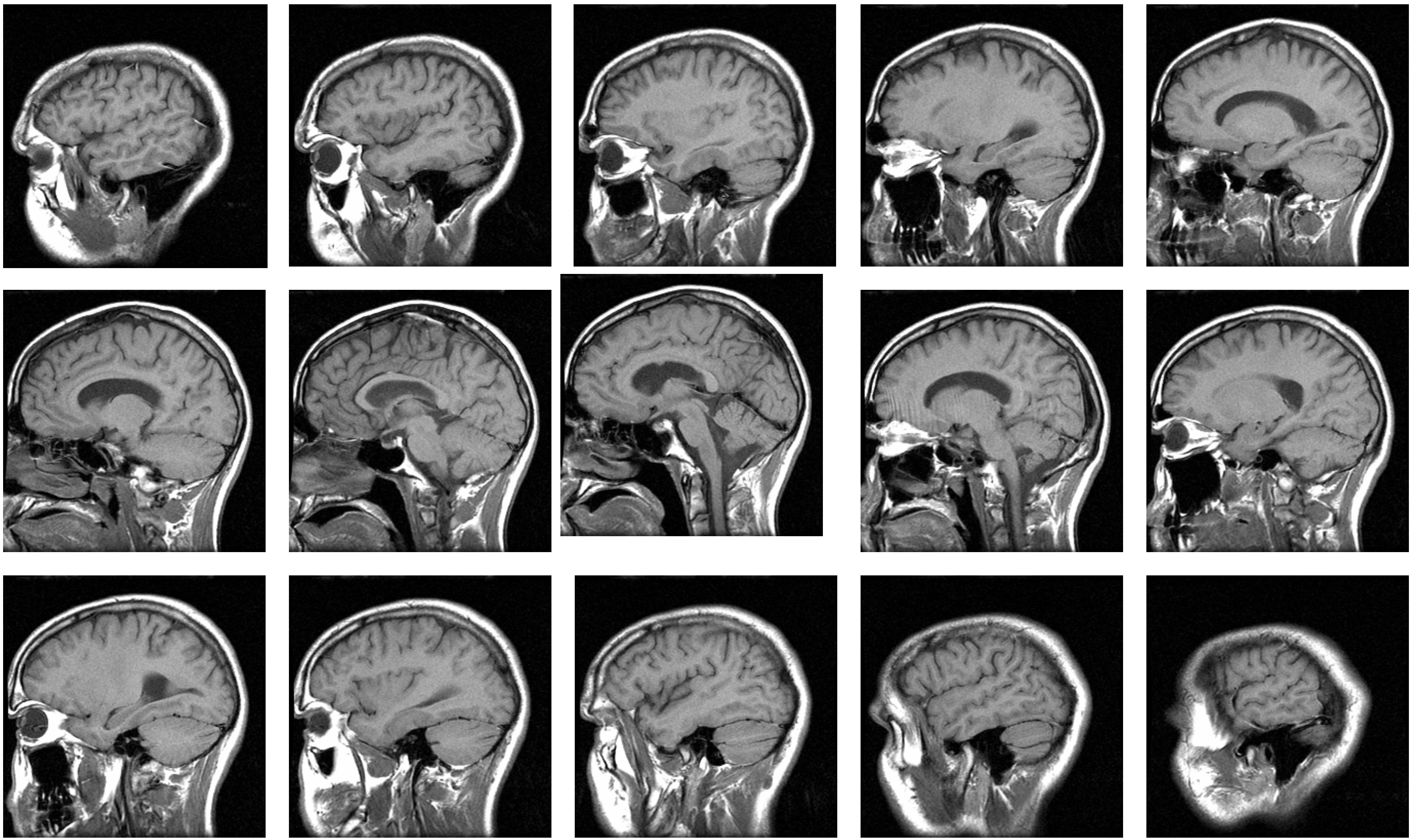
Proton density
T_E = 14 ms
T_R = 1500 ms

T2 weighting SE (double echo)



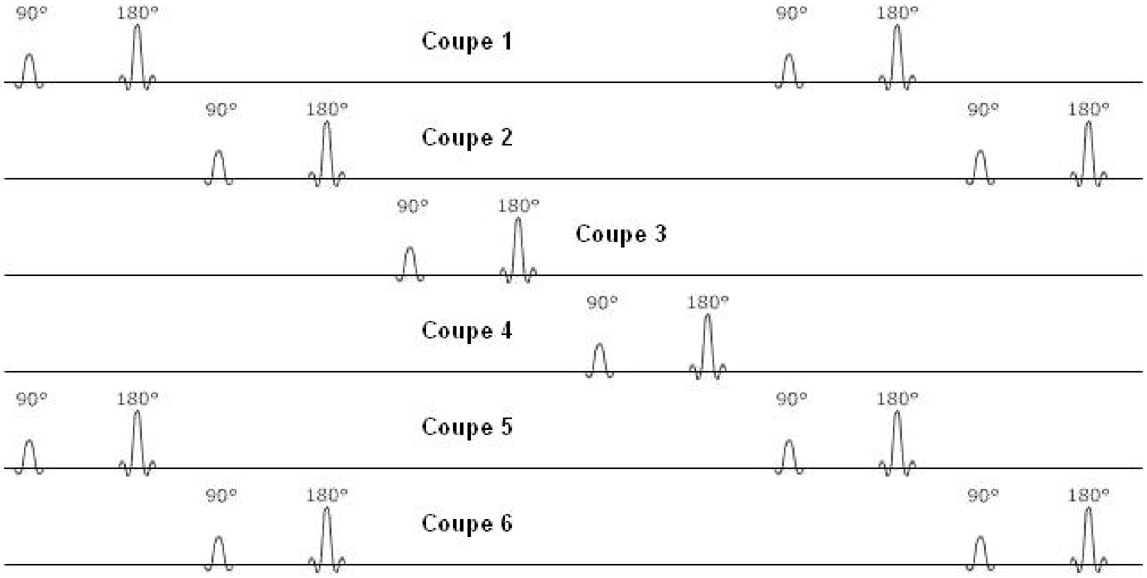


Slice in whatever plane

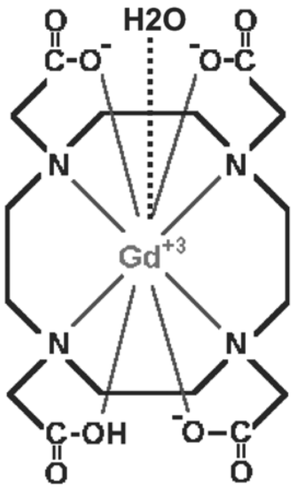


What can we do during TR?

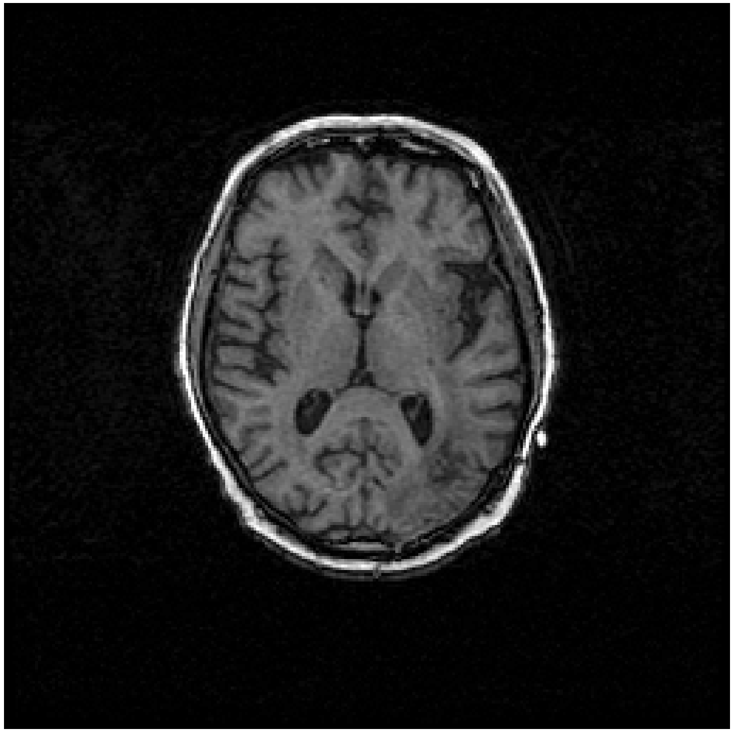
Multi-slice acquisition



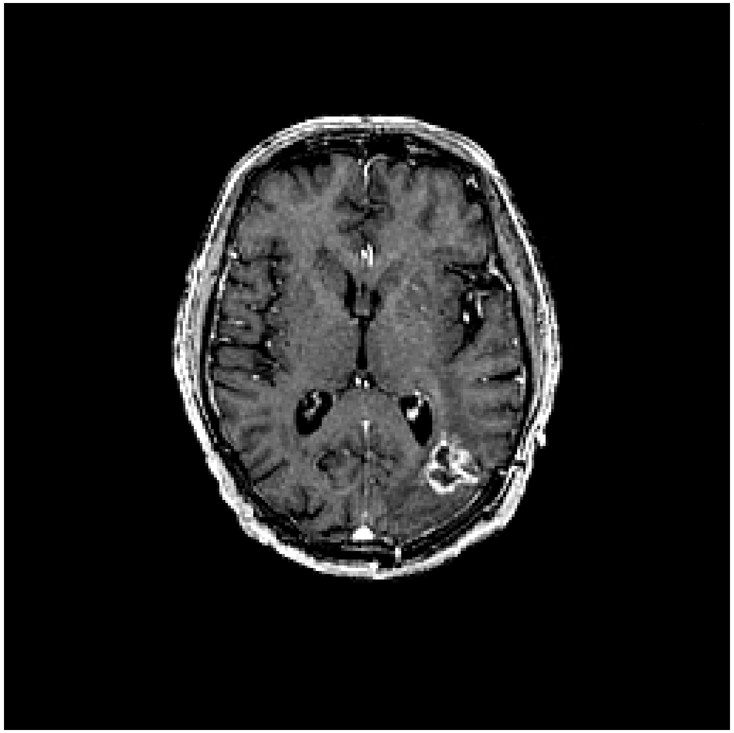
Contrast media : T1 reduction



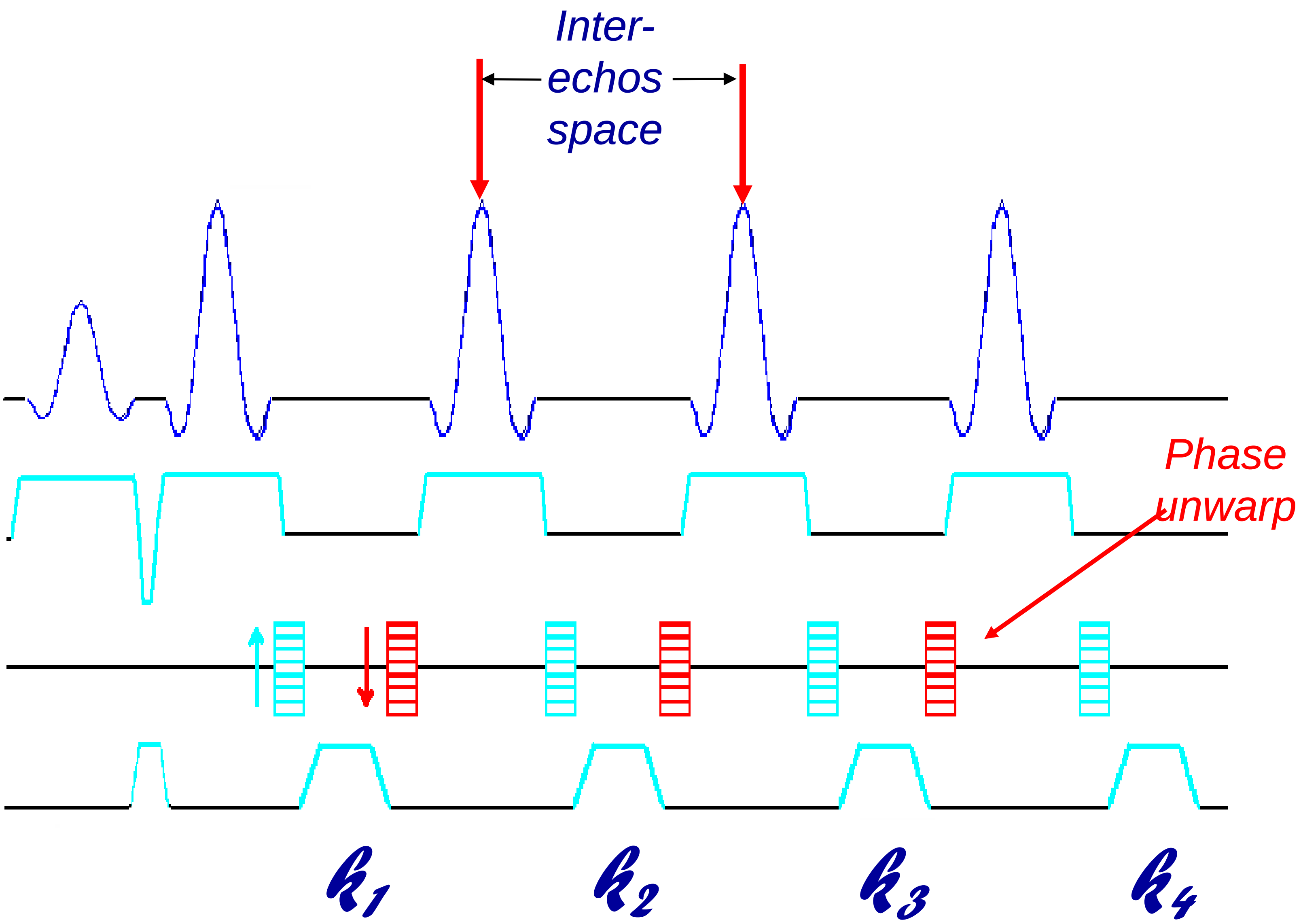
Native T1



T1 + Gd



SE and FSE what differences?

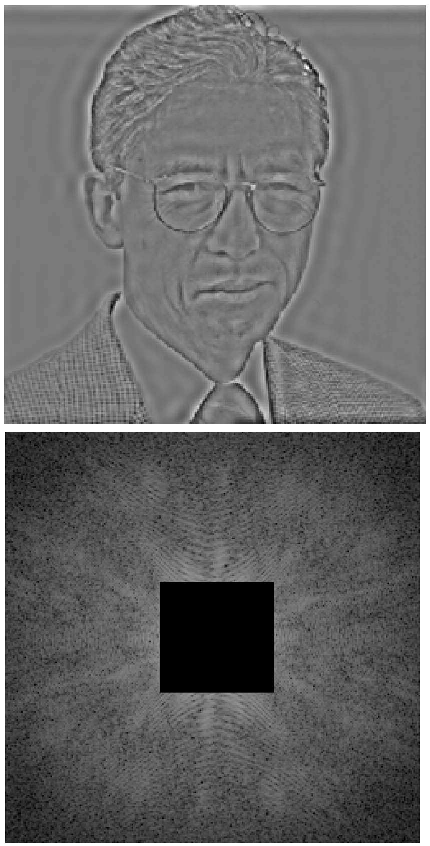


Contrast in FSE

Complete image

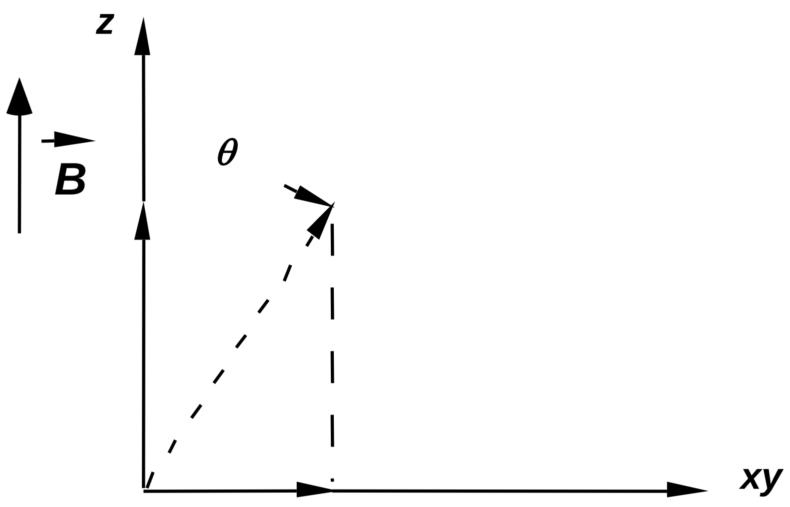
Center of k space

K space periphery



Acquisition time reduction: FSE (T2 contrast)
Other solution : GRE (Gradient Recalled Echo)

- *Principle : lower tip angle for M*

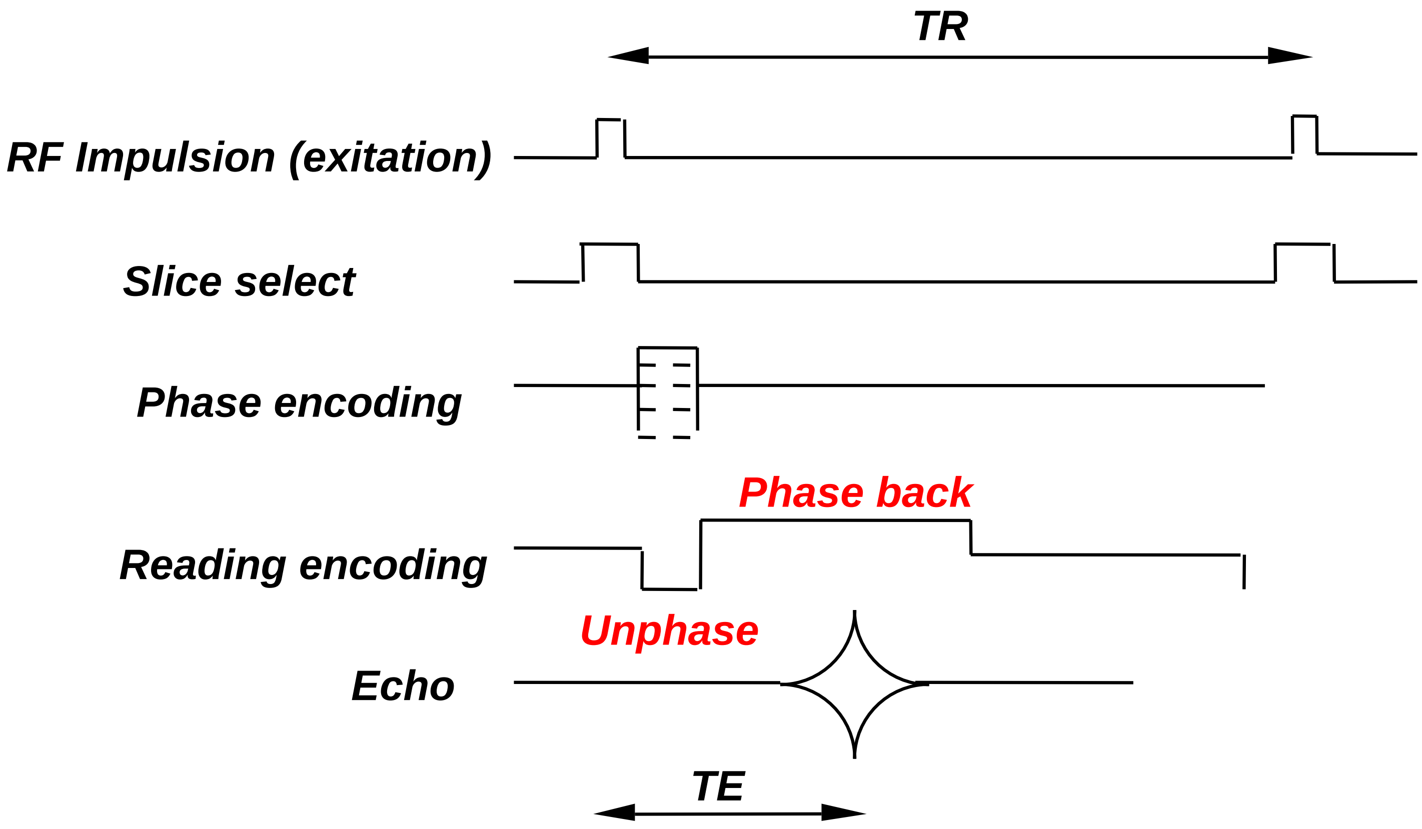


$$M_{xy} = M_o \sin\theta$$

$$M_z = M_o \cos\theta$$

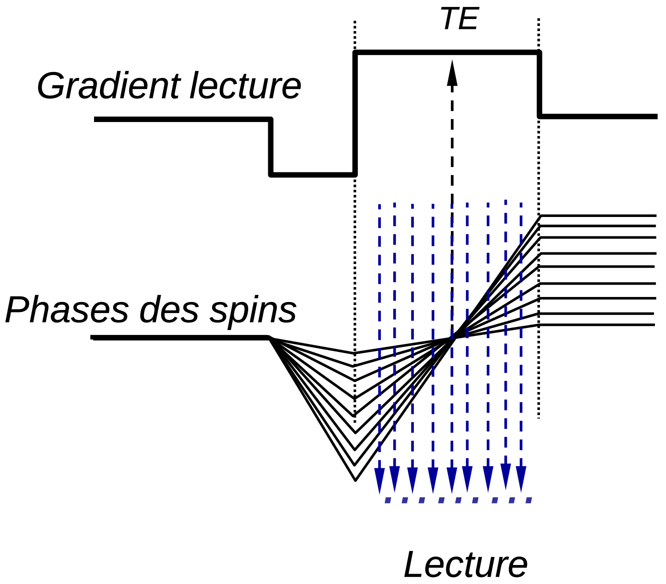
Faster T1 relaxation → TR can be reduced

General GRE sequences



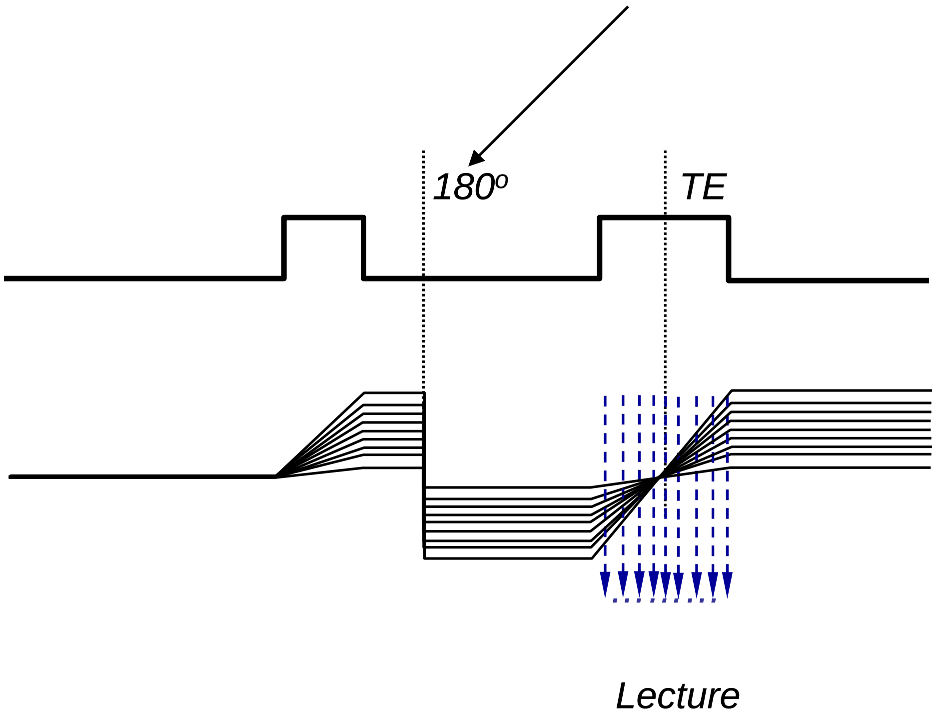
Spin behavior

GRE



ρ, T1, T2*

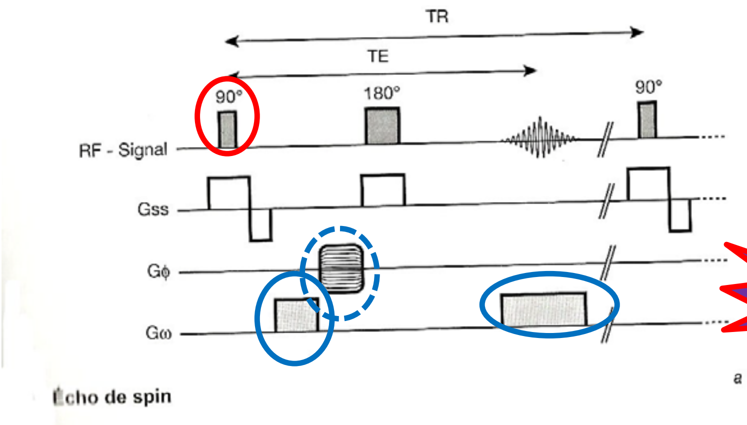
SE with refocalization impulsion

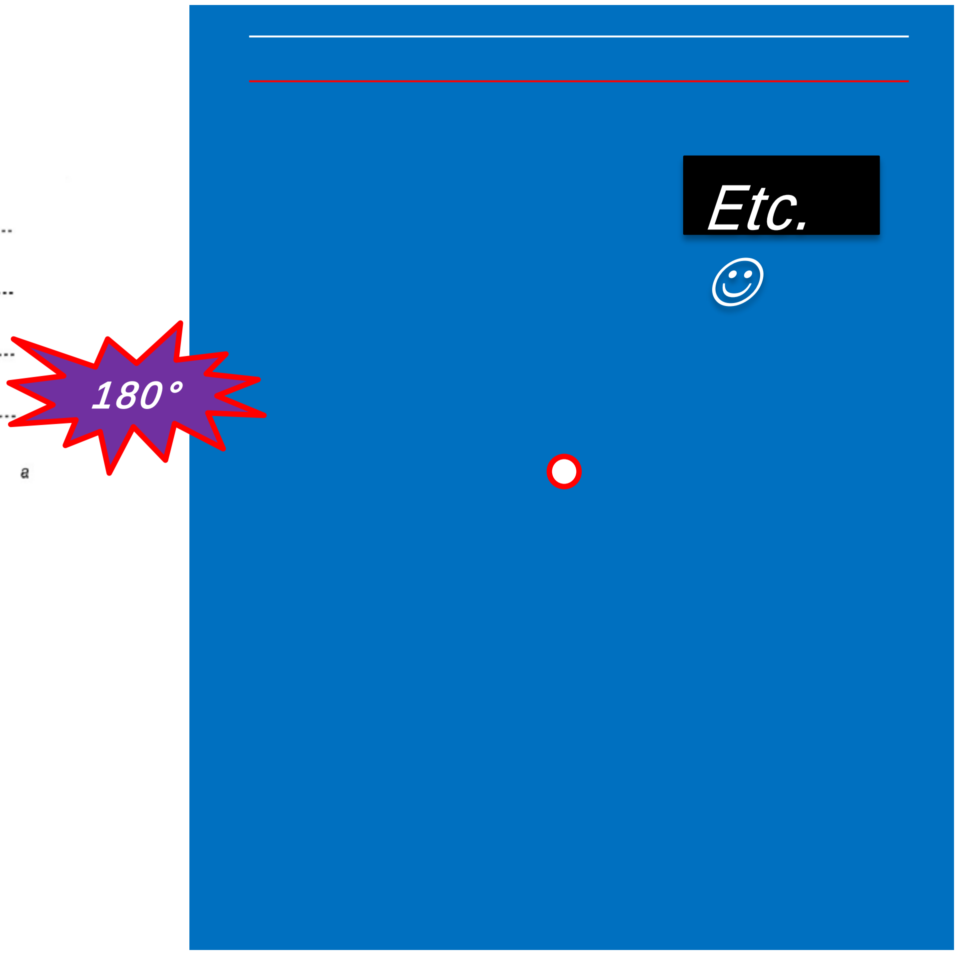


ρ, T1, T2

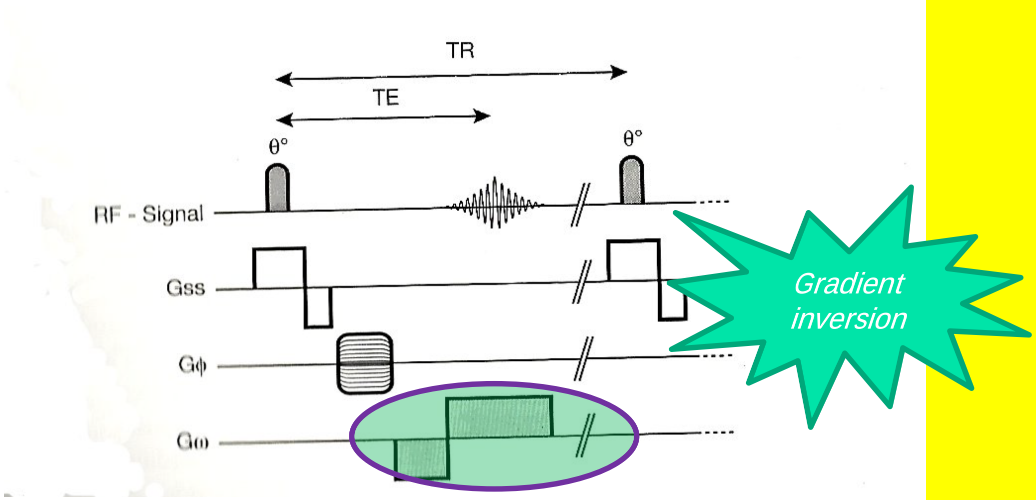
Sequence	TR Short	TR Long	TE Short	TE Long
SE	250–700	> 2000	10–25	> 60
GRE	< 50	> 100	1–5	> 10

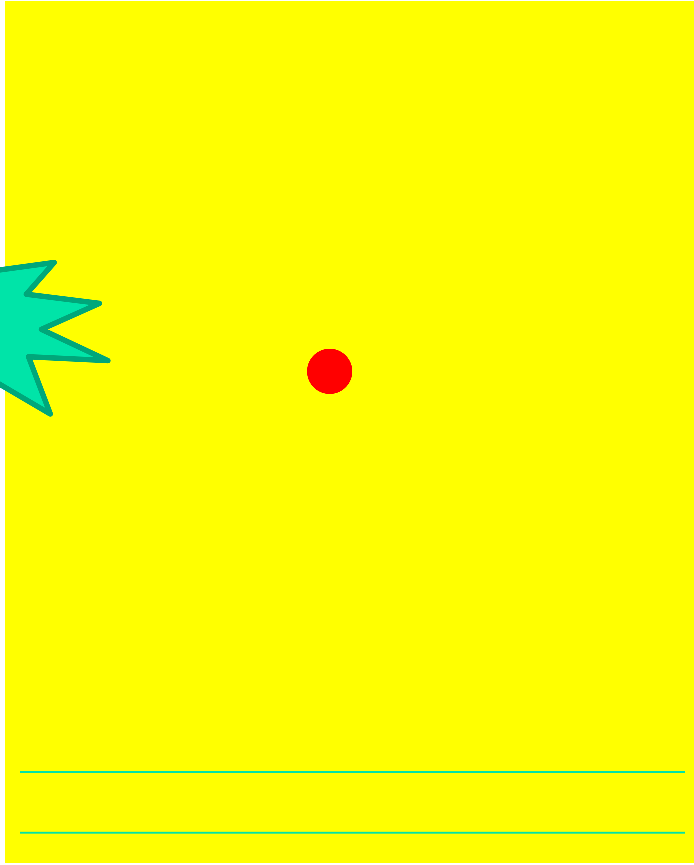
Traveling in k space for spin echo





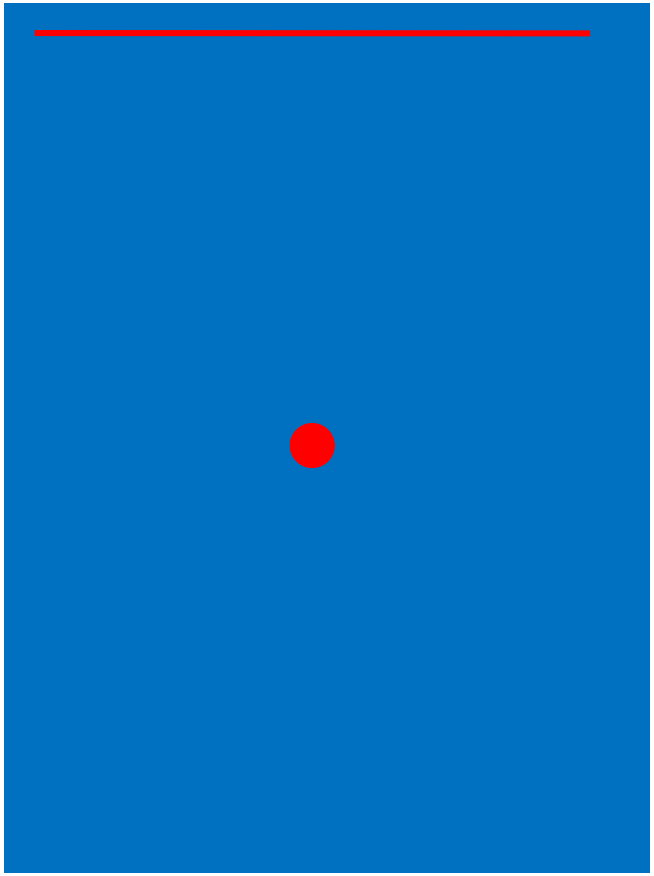
Traveling in k space for gradient echo



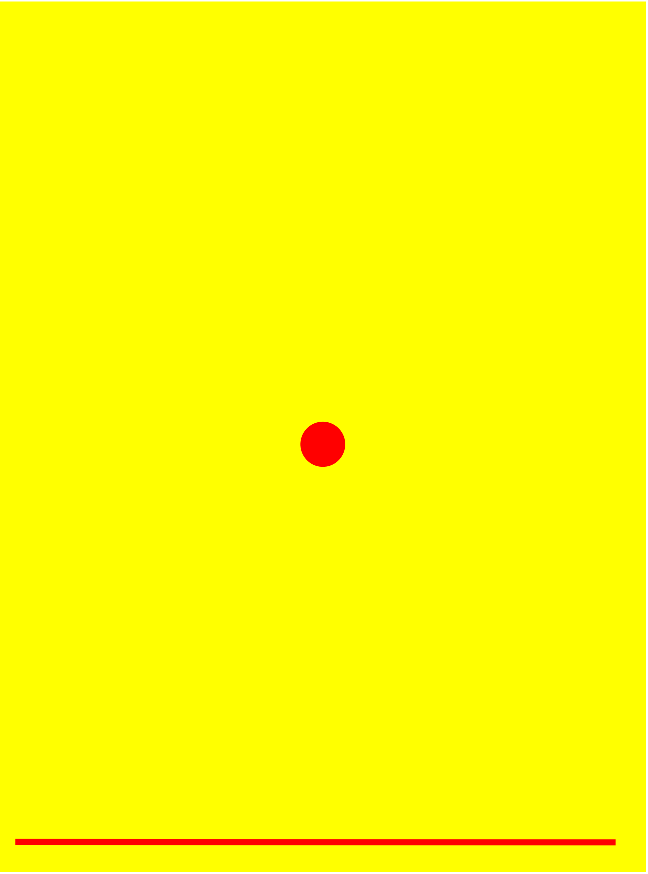


Spin echo versus gradient echo

SE

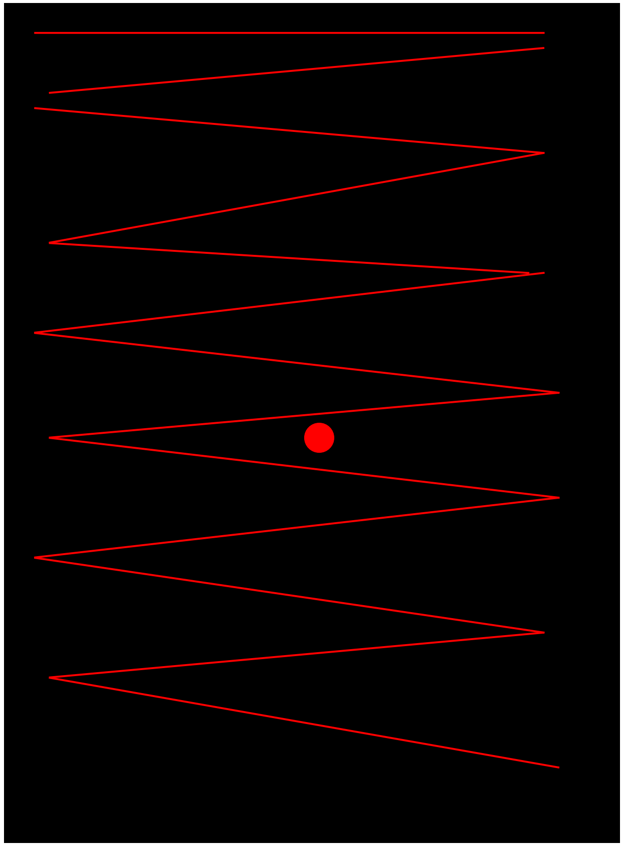


GRE



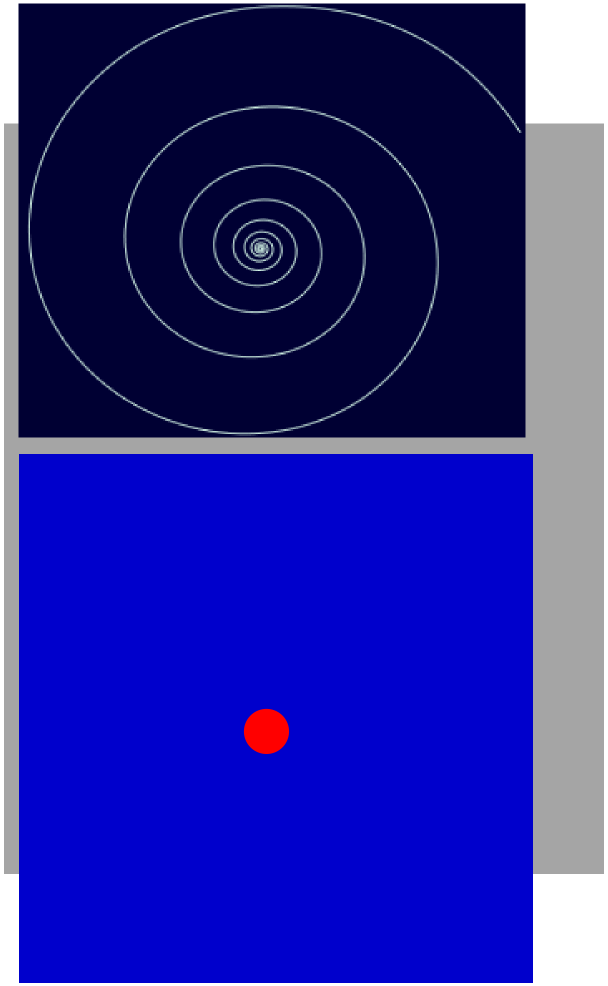
Traveling in k space for echo planar

- Excitation 90°
- Continuous G_y
- G_x alternates between positive and negative values



Traveling in k space for spiral data sampling

- Sinusoidal variation of G_y and G_x
- Variable intensity
- For what ?



Functional imaging: MR spectroscopy