

Module Medical Imaging: Lecture 1

(Part: Schick/Machann/Klose)

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Section of Experimental Radiology
Department of Diagnostic and
Interventional Radiology

2018





Computed tomography
(CT)
→ X-rays



Magnetic Resonance tomography
(MT)
→ Strong magnetic field

CT

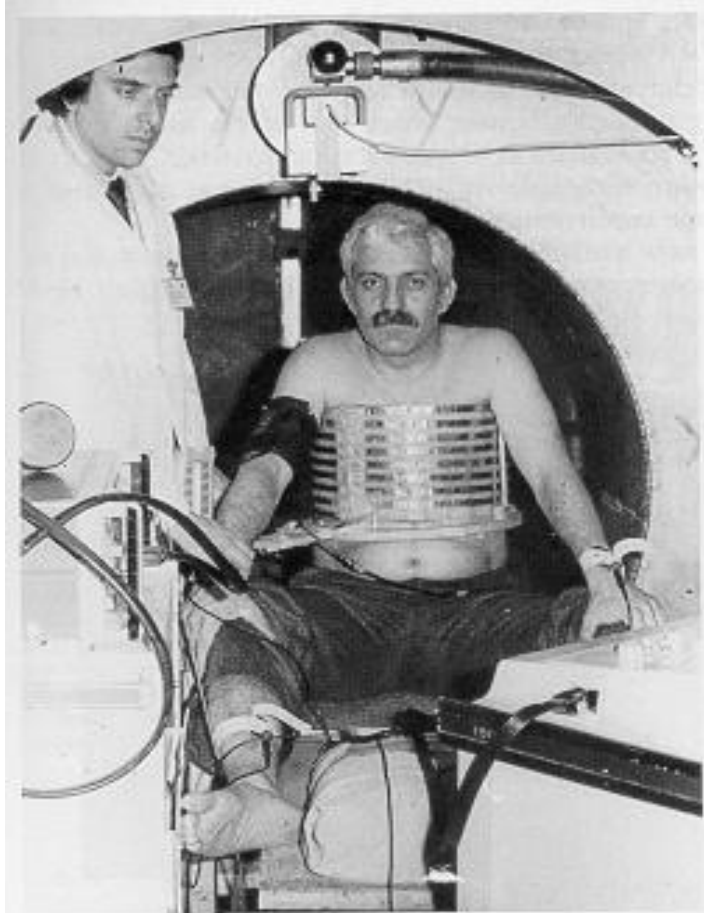


MR

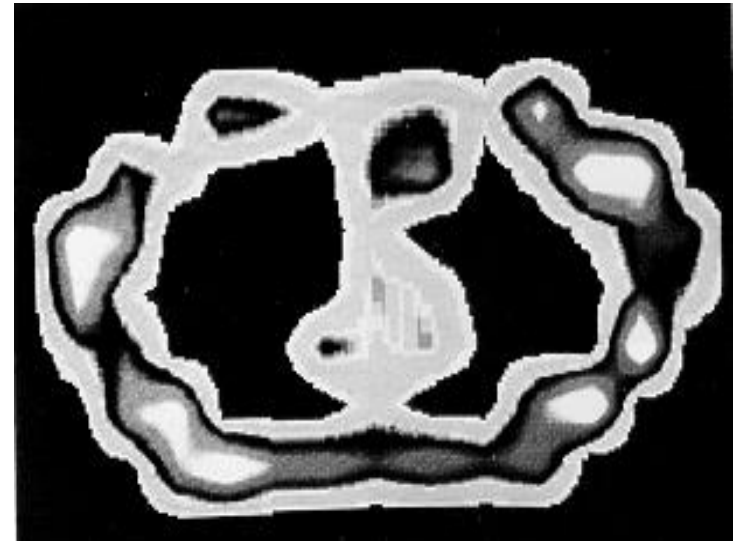


History of MRI

First examination of a human subject in 1977



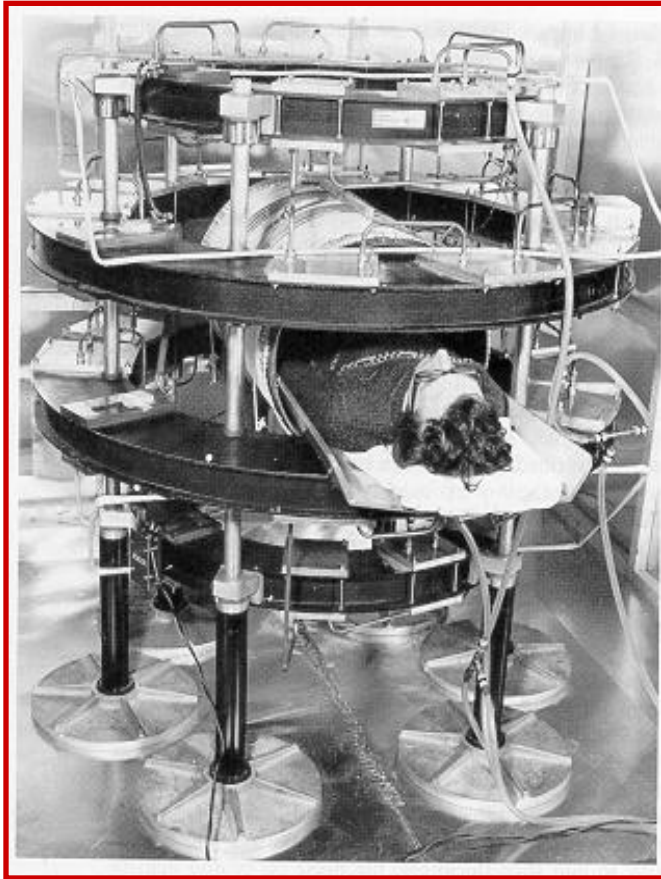
volunteer:
Raymond Damadian



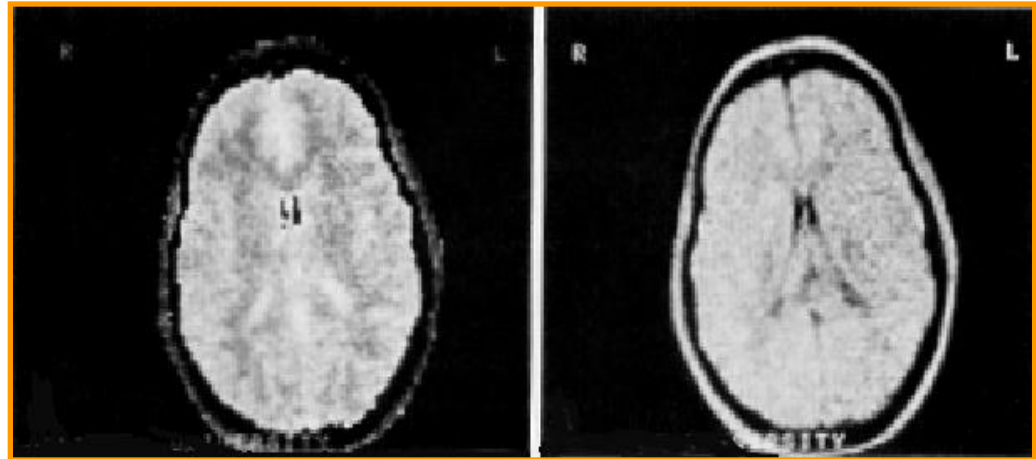
resulting MR image
recorded in a measuring
time longer than 1 hour

History of MRI

State of the art 1981



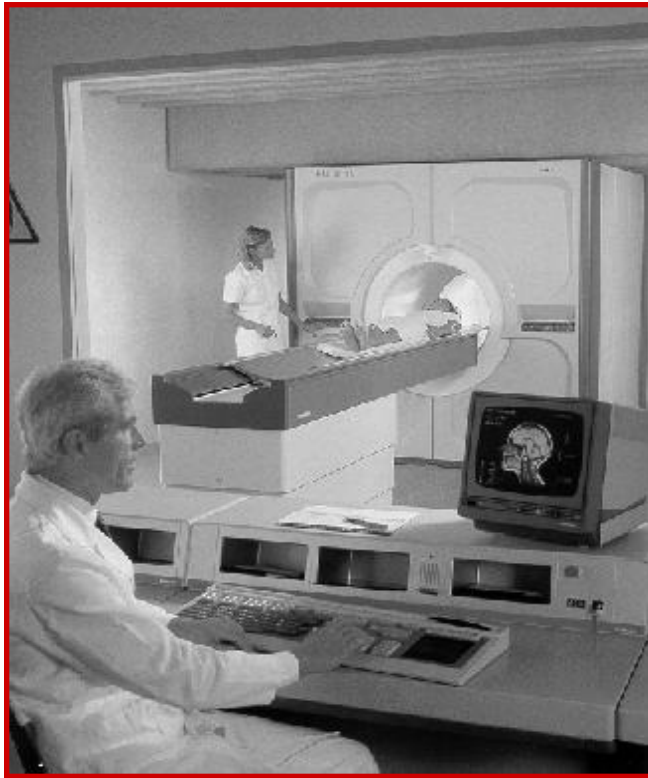
MR-imager, Aberdeen 1981



MR-images from Aberdeen 1981
measuring time about 45 min.

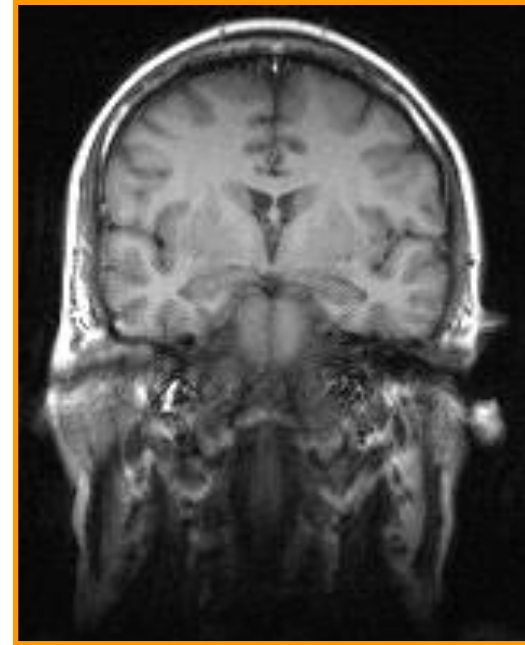
History of MRI

State of the art 1985



MR-imager
Siemens 1985

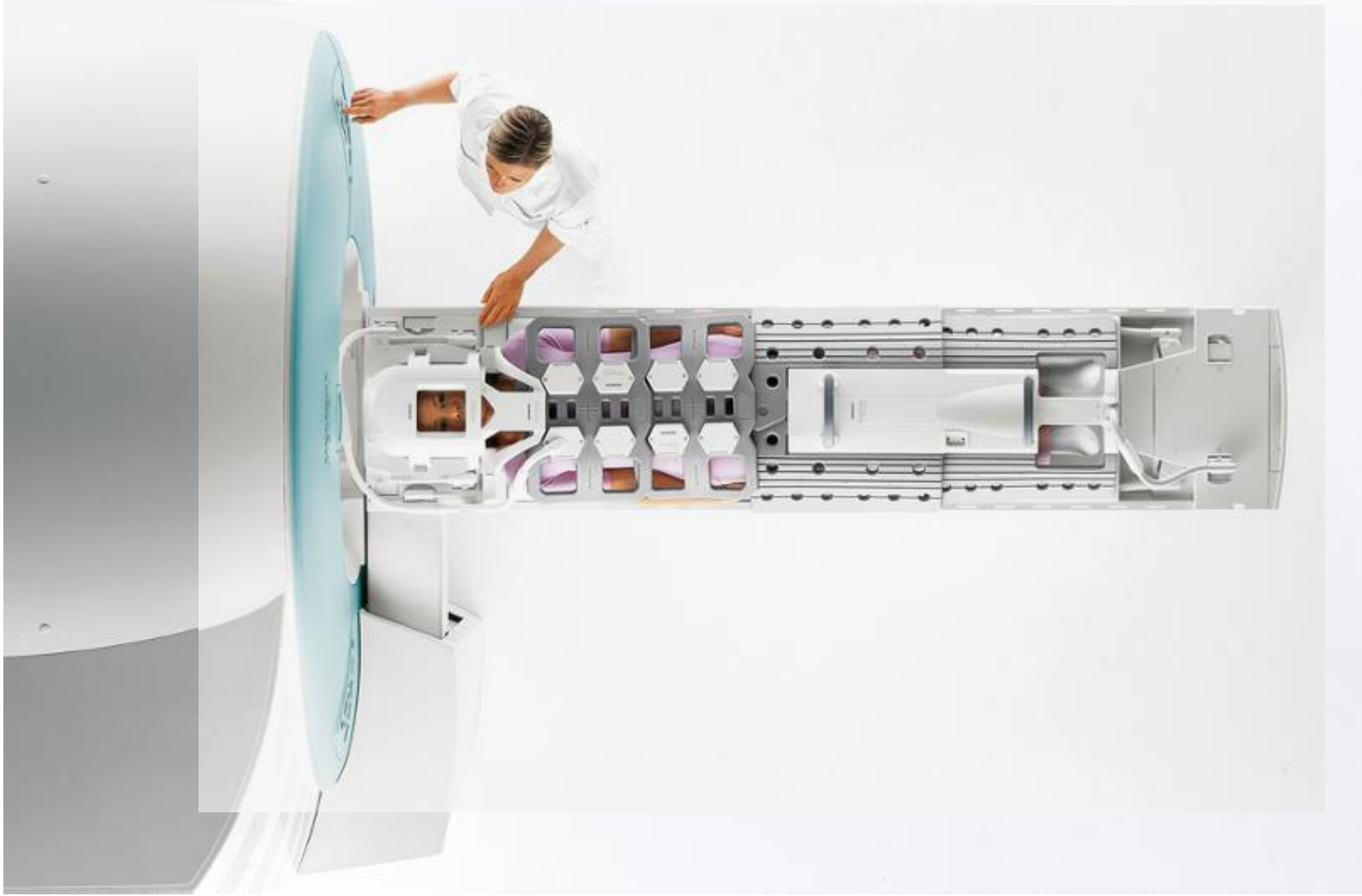
gradient strength 3 mT/m



First examples after
installation in Tübingen
measuring time about 5 min.

History of MRI

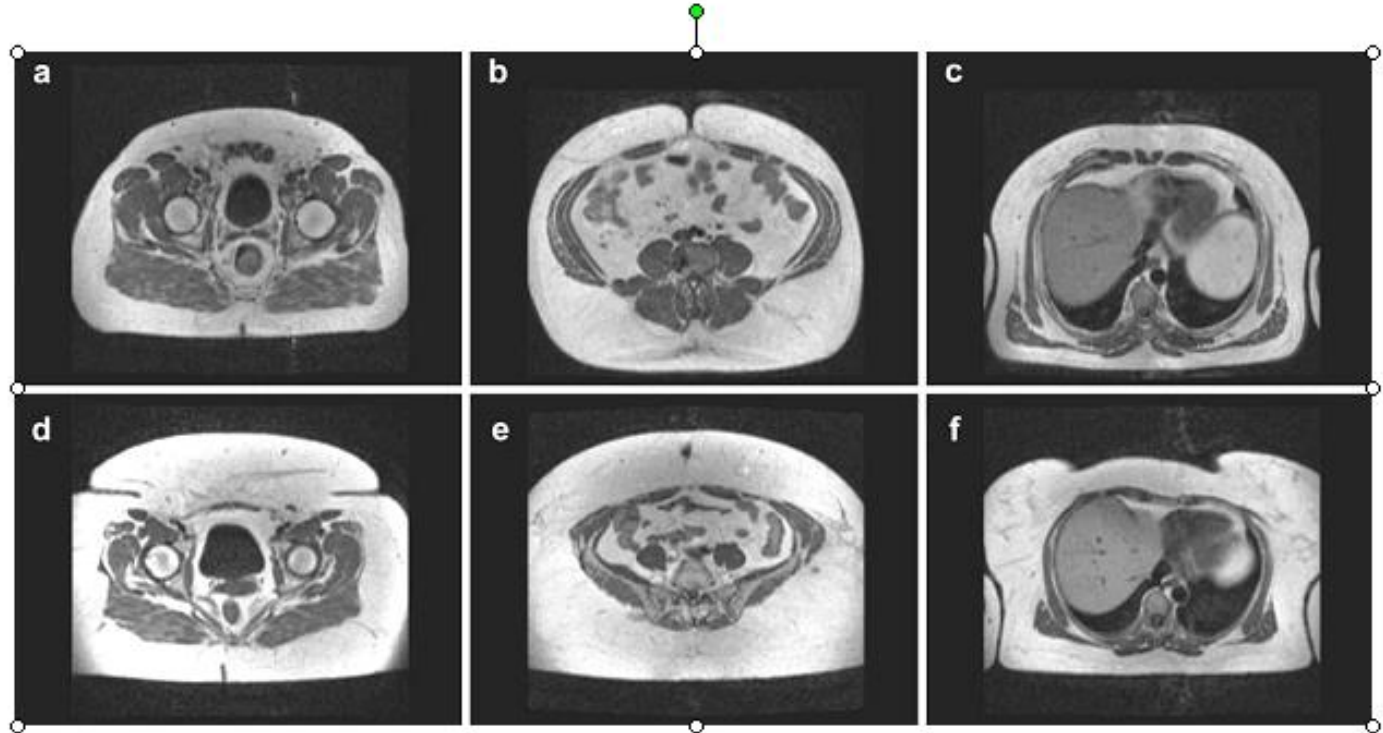
State of the art 2005



History of MRI

State of the art 2007

BMI 40.8



BMI 52.8

Springer F et al., Obes Surg. 2010 Jun;20(6):749-56.



History of MRI

State of the art 2012

g throughput)

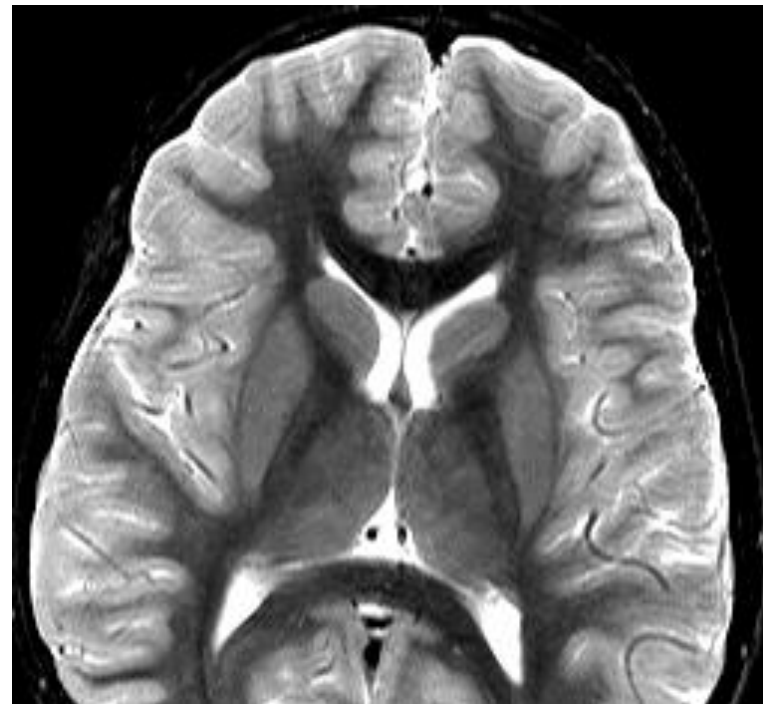


History of MRI

Typical examination of a human subject today



commercial clinical
MRI system in
Tübingen 2015
(Siemens, 1.5 Tesla)



resulting MR image
of the brain recorded
in a measuring time of
less than 1 minute

History of MRI

Nobel Prize in Medicine 2003 for “their discoveries concerning magnetic resonance imaging”



Peter Mansfield

Paul C. Lauterbur

Outline

- **Basic principles of magnetic resonance:**
spins, magnetization, RF excitation, relaxation
- **Standard contrasts:**
proton density, T1 and T2 weighting
- **Hardware components of an MR imaging system:**
static magnetic field, gradient fields, RF field T/R
- **Cartesian spatial encoding:**
2D and 3D encoding
- **Standard sequences:**
gradient echo, spin echo, and fast spin echo

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Sources of the MR Signal

In **medical imaging** almost always signals from ^1H **nuclei** are detected („proton imaging“)

Only relatively small and movable molecules containing hydrogen contribute to the signal.

Actually only the signals of **two types of molecules** occurring in large quantities in the body are visible in MR images:

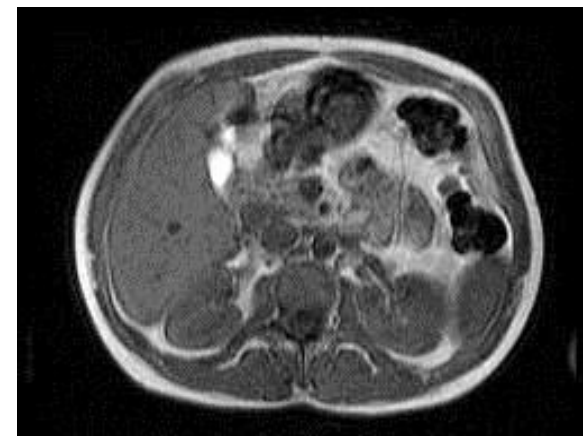
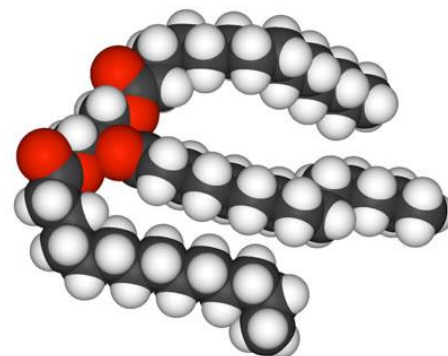
Water



- in the cytosol of the cells
- in the interstitial space
- in body fluids

Fat/Lipids

- as tri(acyl)glycerol in the fat vacuoles of the adipocytes



Signal components of other molecules containing ^1H are below the noise floor (e.g. glucose, amino acids...)

Sources of the MR Signal

nuclei with odd mass number:

$I = 1/2, 3/2, 5/2, \dots$ (s.u.)

nuclei with odd proton- and neutron-number:

$I = 1, 2, \dots$ (^2H , ^{14}N)

nuclei with even proton- and neutron-number:

$I = 0$ (^{12}C , ^{16}O)

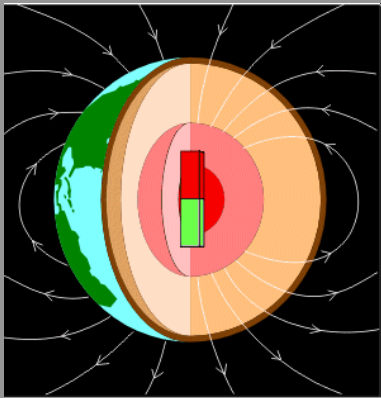
MR-properties of different nuclei

nucleus	spin	natural occurrence	Larmor frequency (MHz / 1 T)	relative sensitivity*
^1H	1/2	99,984%	42,58	1
^{13}C	1/2	1,108%	10,71	$1,76 \times 10^{-4}$
^{19}F	1/2	100%	40,05	0,833
^{23}Na	3/2	100%	11,26	$9,27 \times 10^{-2}$
^{31}P	1/2	100%	17,23	$6,65 \times 10^{-2}$

* Calculated by the product of MR-sensitivity (proportional to $\gamma^3 \cdot I(I+1)$) and the natural occurrence

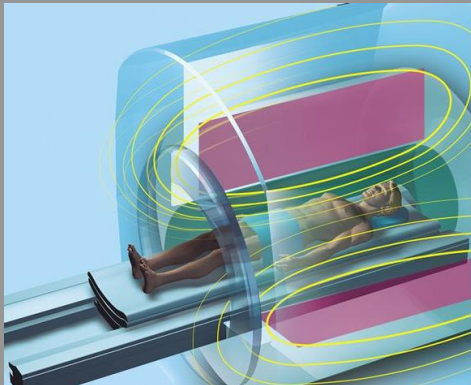
Principle of Magnetic Resonance

What is necessary for generating detectable signals by atomic nuclei?



Magnetic field of the earth: approx. $30 \mu\text{T}$

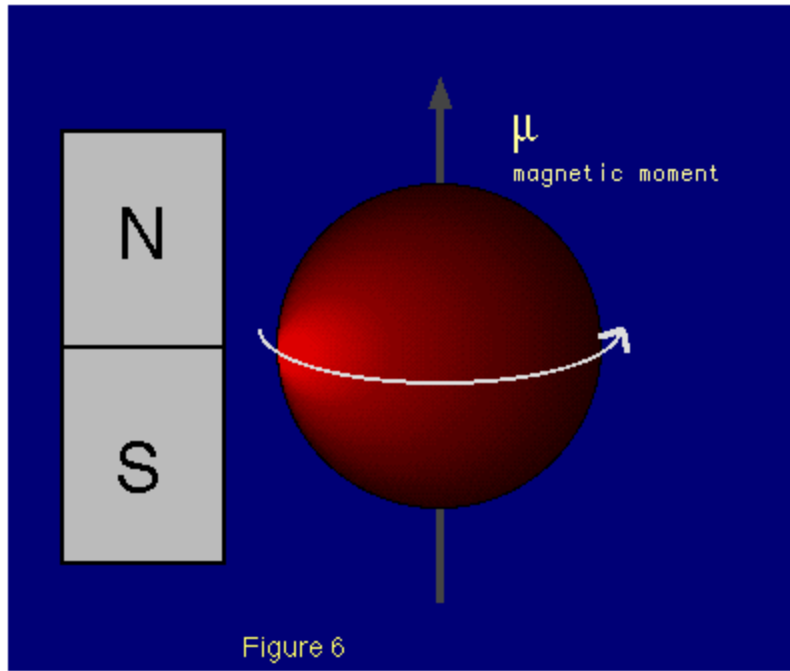
We need a strong magnetic field!



MR-imagers: $1,5 - 7 \text{ T}$

Principle of Magnetic Resonance

nuclear spin and magnetic moment

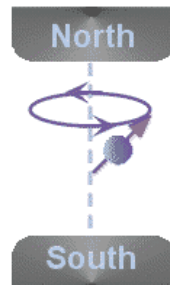


mechanical
property:

spin $\vec{I}$

magnetic
property:

magnetic
moment μ

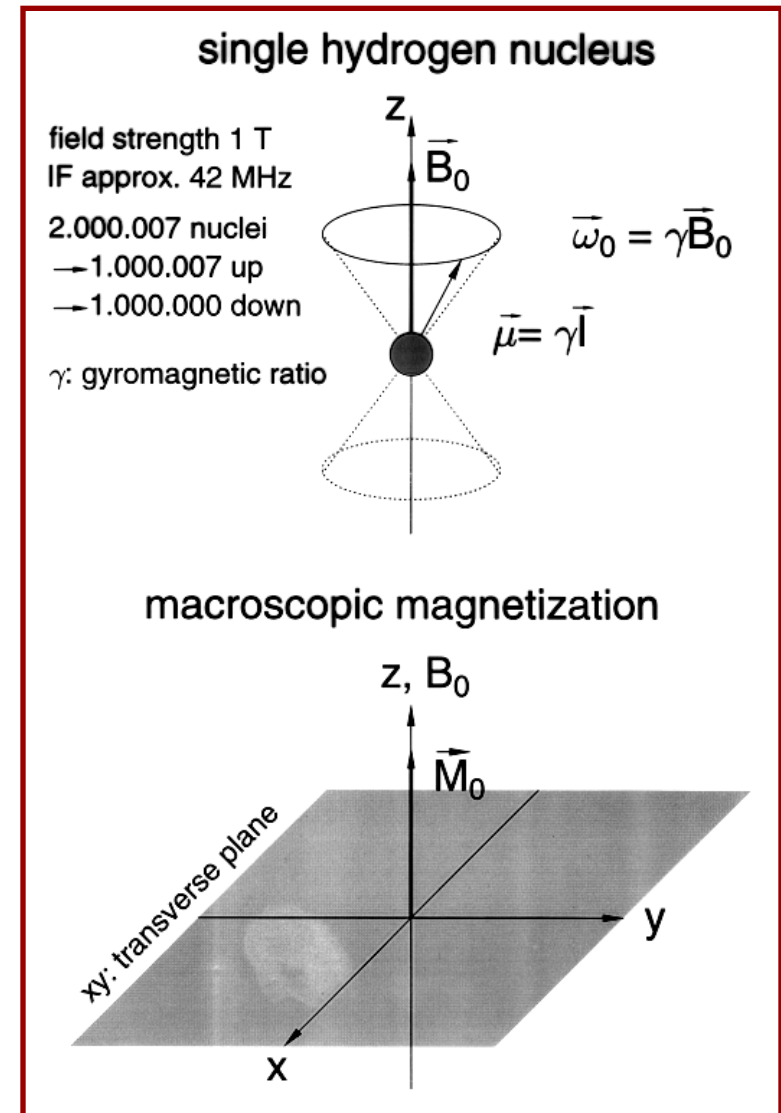


Principle of Magnetic Resonance

Single nucleus in
the magnetic field

and

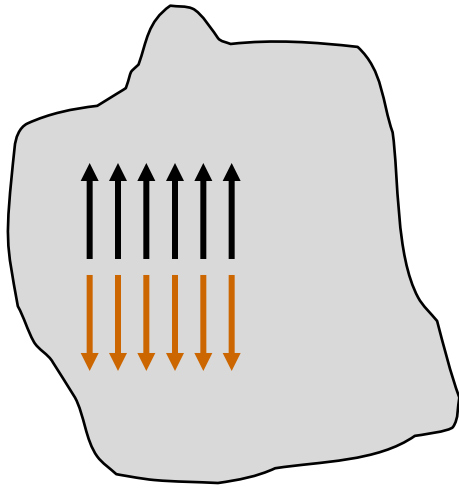
macroscopic
magnetization M_0



Principle of Magnetic Resonance

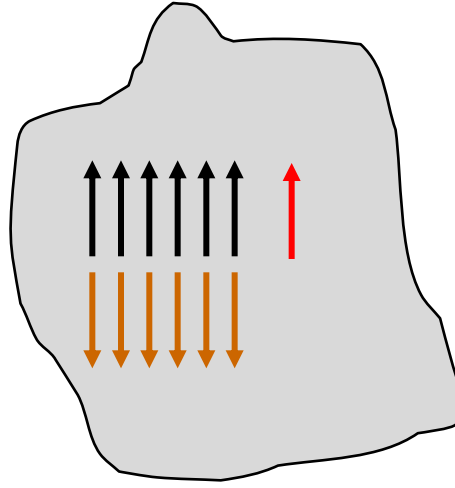
Effect of increasing field strength B_0

0 Tesla



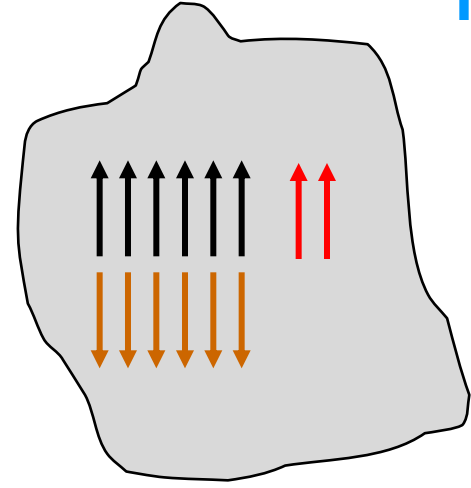
0 : 1.000.000

1.5 Tesla



5 : 1.000.000

3 Tesla



10 : 1.000.000

↑ B_0

Surplus of nuclei with spin in the direction of the applied magnetic field B_0 at room temperature

Principle of Magnetic Resonance

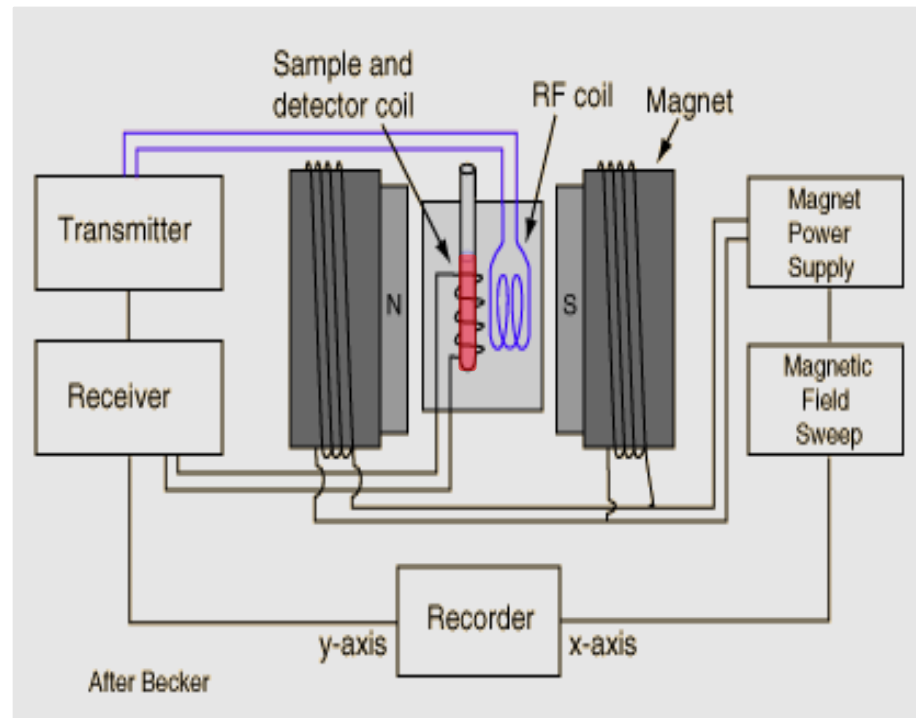
Nuclear Magnetic Resonance (NMR) – Experiments with diamagnetic liquids (1946)



*Edward Mills
Purcell
(1912-1997)*



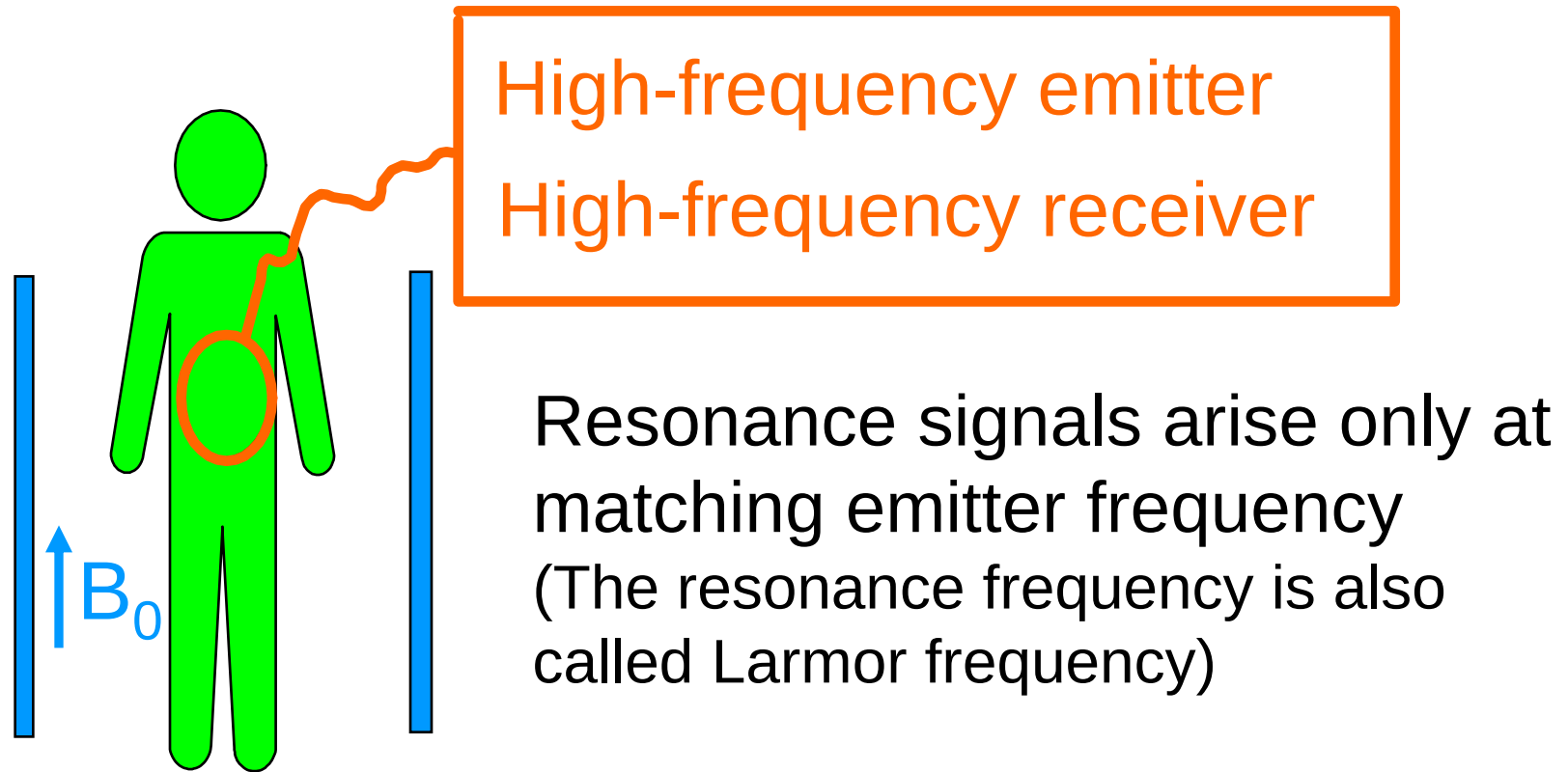
*Felix Bloch
(1905-1983)*



*Both scientists detected
NMR signals in liquids
(Nobel prize 1952)*

Principle of Magnetic Resonance

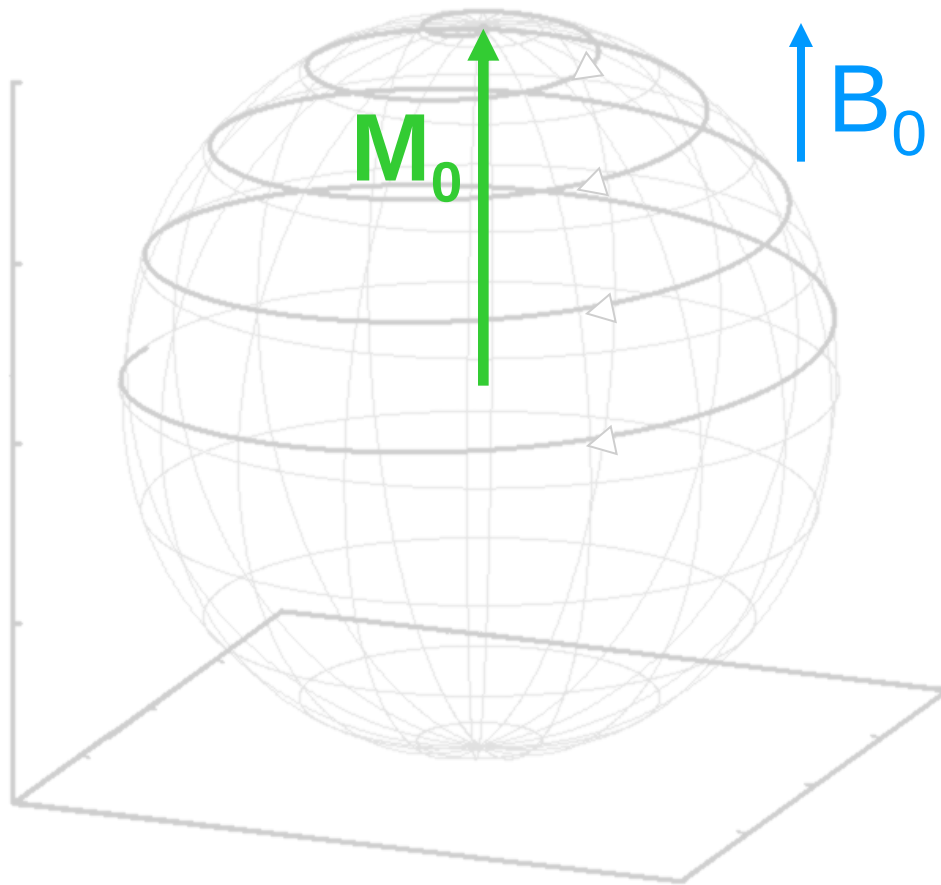
Generation of MR signals



42.6 MHz per Tesla for ^1H (hydrogen)

Principle of Magnetic Resonance

Concept of nuclear magnetization as a vector



Magnetization M is a vector in space.

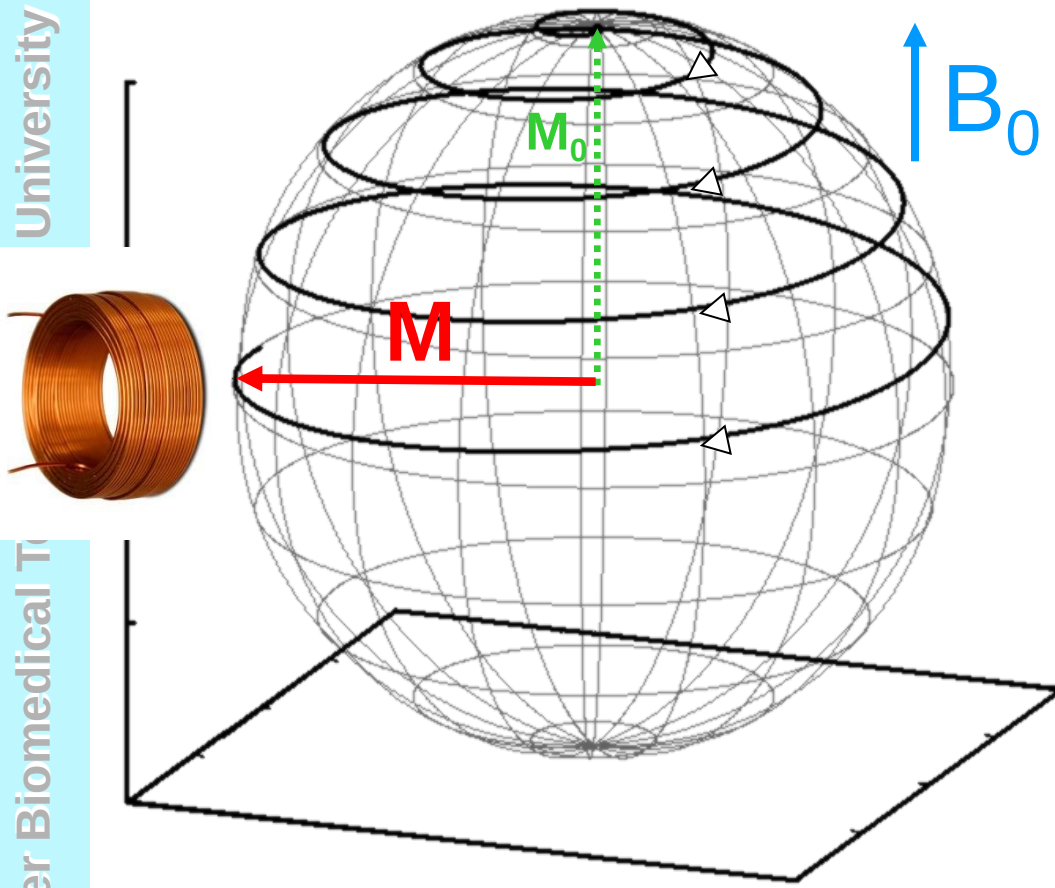
M represents the magnetic effect of many identical nuclei within a volume.

In the **equilibrium state**, **M_0** is aligned parallel to the static magnetic field **B_0** .

The behaviour of **M** follows simple laws as described by the **Bloch equations**.

Principle of Magnetic Resonance

RF excitation of magnetization (90°-pulse)

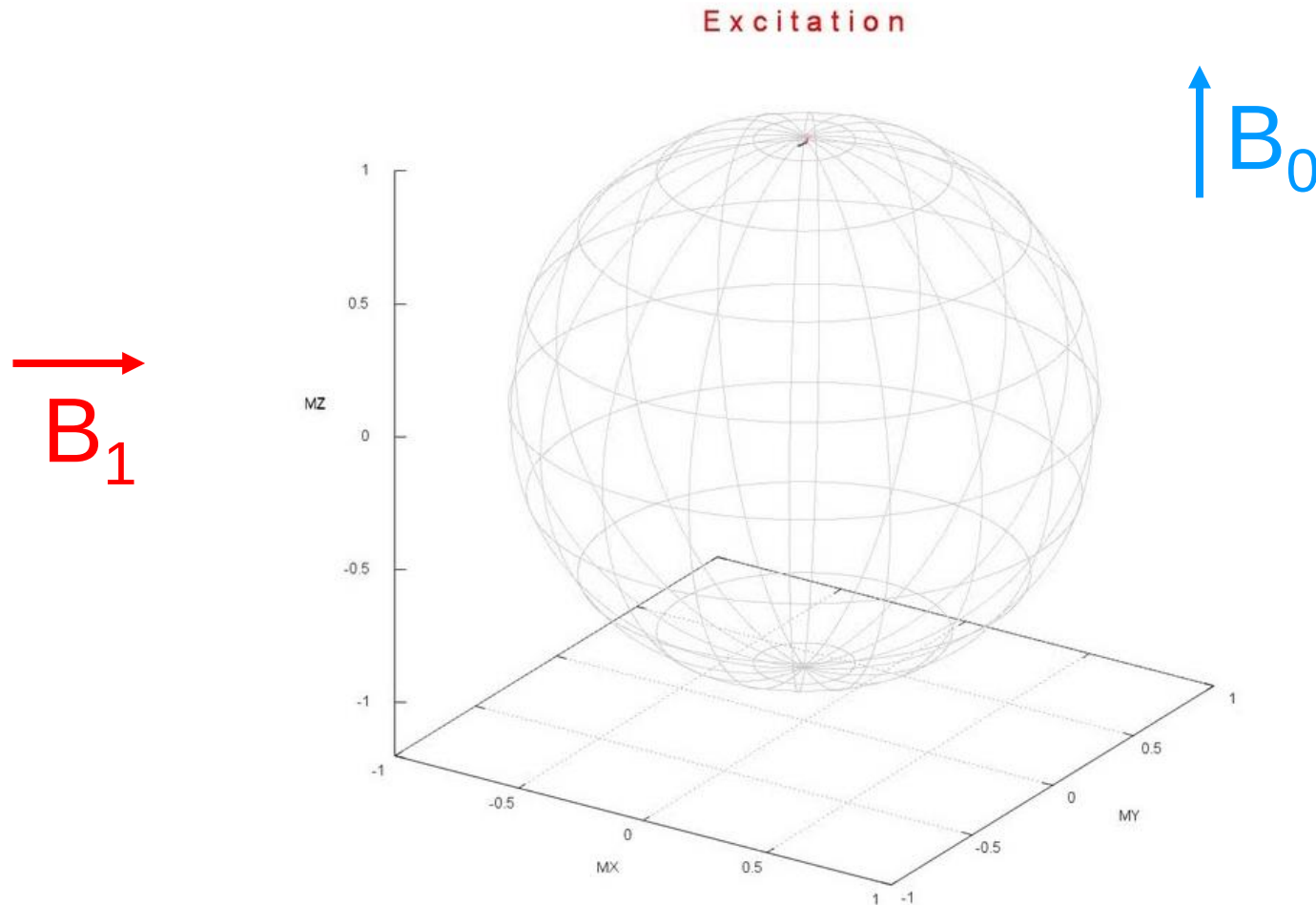


Duration of excitation is approx. 1-5 ms

Thereby a **rotating transverse magnetization** is generated which induces an electrical voltage in a receiver coil.

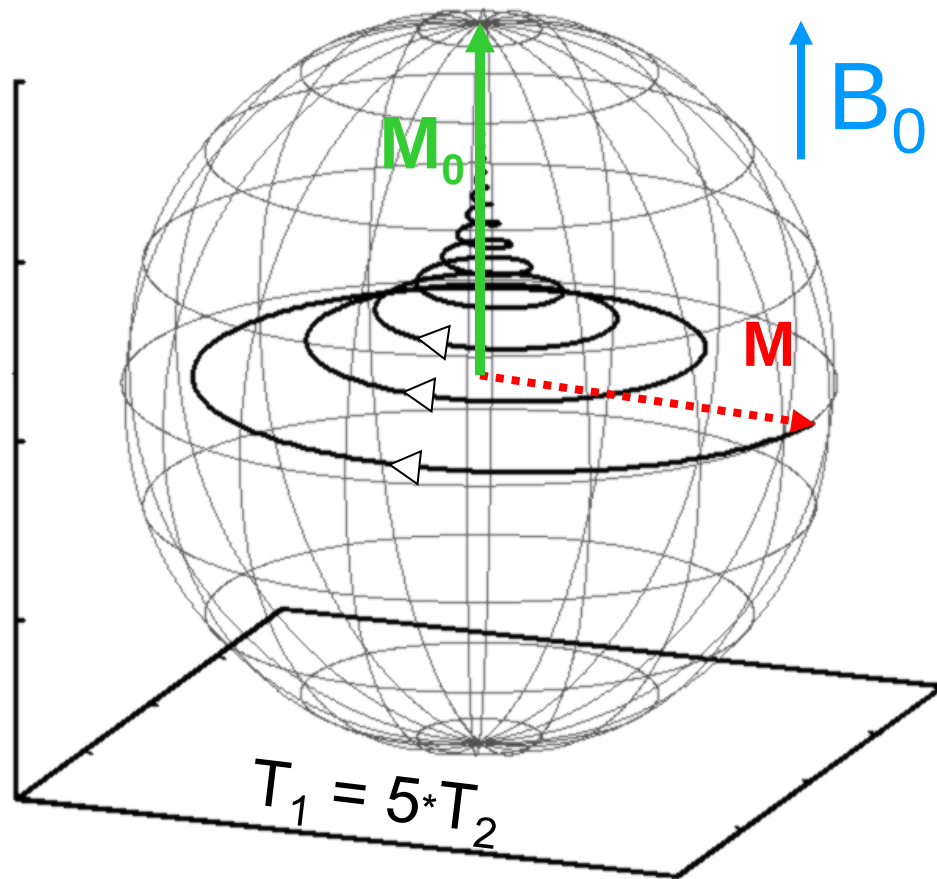
Animation: RF Excitation

The trajectory of the arrowhead of the magnetization vector:



Principle of Magnetic Resonance

Relaxation of magnetization (following 90°-pulse)

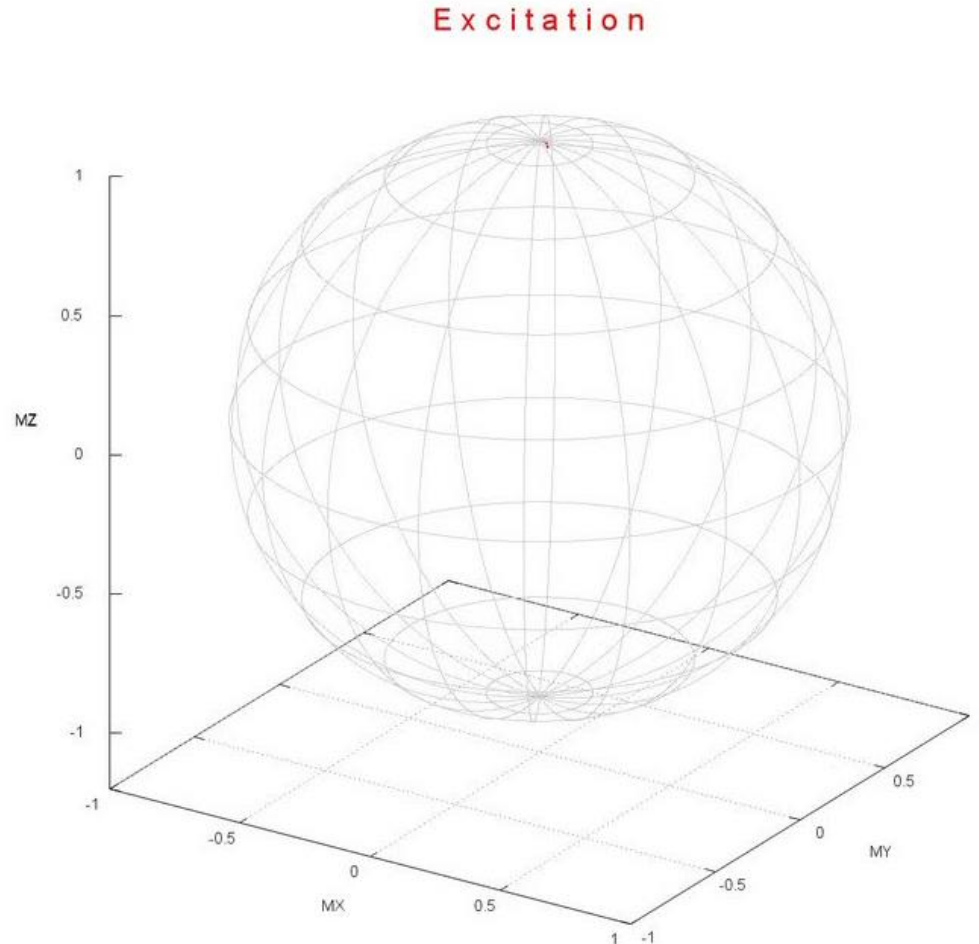


Duration of relaxation in tissue: several seconds

Relaxation leads the magnetization back to the **equilibrium state** M_0

Animation: RF Excitation + Relaxation

The trajectory of the arrowhead of the magnetization vector:

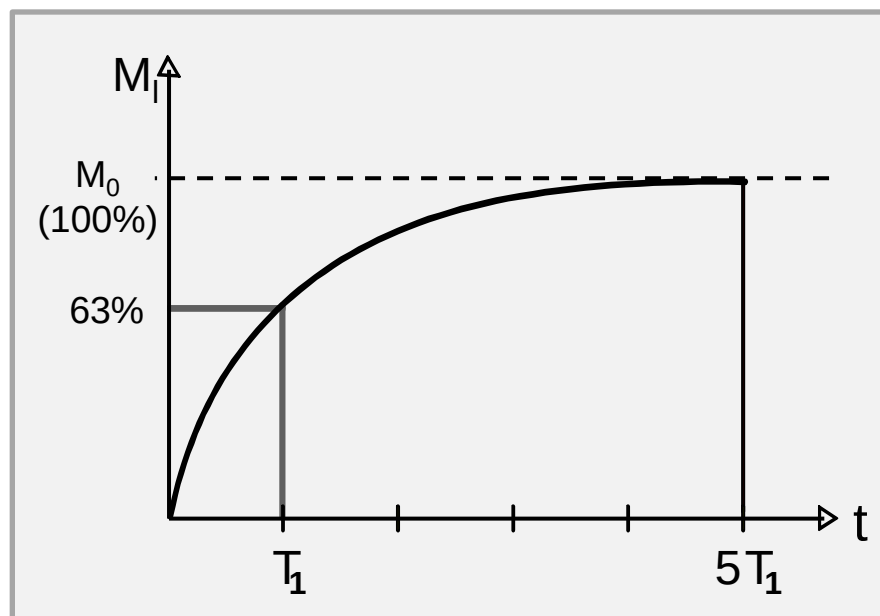


Relaxation T1, T2

Behaviour of the rotating nuclear magnetization

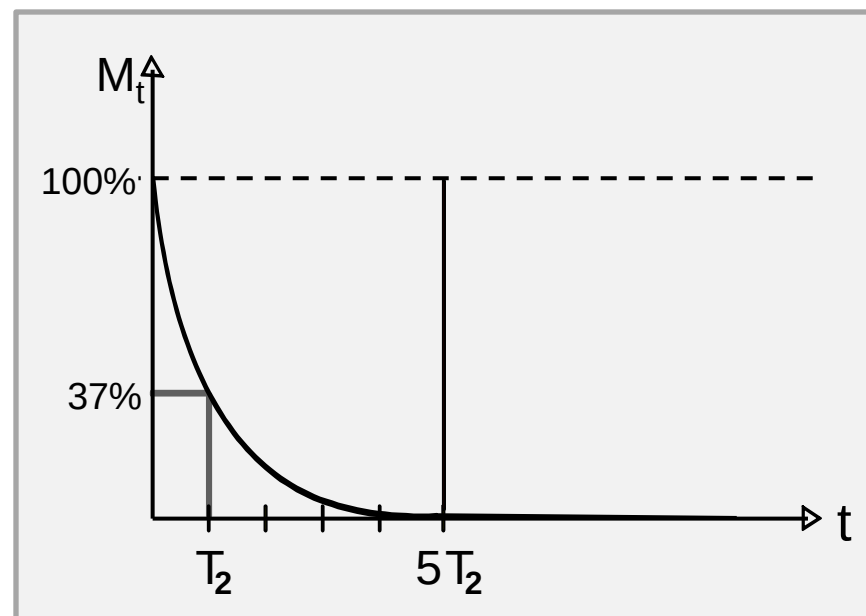
Relaxation following a 90° excitation:

Exponential increase of the longitudinal magnetization M_l



$$M_z(t) = M_0 - (M_0 - M_z(t_0)) \cdot \exp(-t/T_1)$$

Exponential decrease of the transverse magnetization M_t



$$M_t(t) = M_t(t_0) \cdot \exp(-t/T_2)$$

Relaxation T1,T2

Important definitions:

Relaxation:

Relaxation is the tendency of the nuclear magnetization vector to regain the equilibrium state. Relaxation occurs when a specimen is newly inserted to a magnetic field or after a disturbance of the magnetization vector (e.g. by an RF-pulse).

T1: Longitudinal relaxation time

During the time period T1 the distance between the longitudinal magnetization and equilibrium value M_0 decreases to $1/e$ (37%) compared to the initial distance.

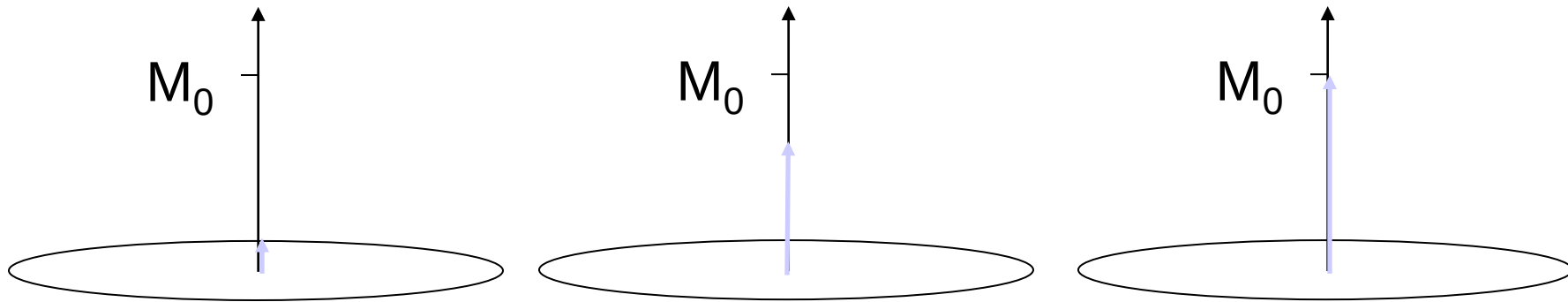
T2: Transversal relaxation time

During the time period T2 the transverse magnetization decreases to $1/e$ (37%) of the initial value.

Relaxation T1,T2

Exponential longitudinal relaxation with T1:

$$M_z(t) = M_0 - (M_0 - M_z(t_0)) \cdot \exp(-t/T_1)$$

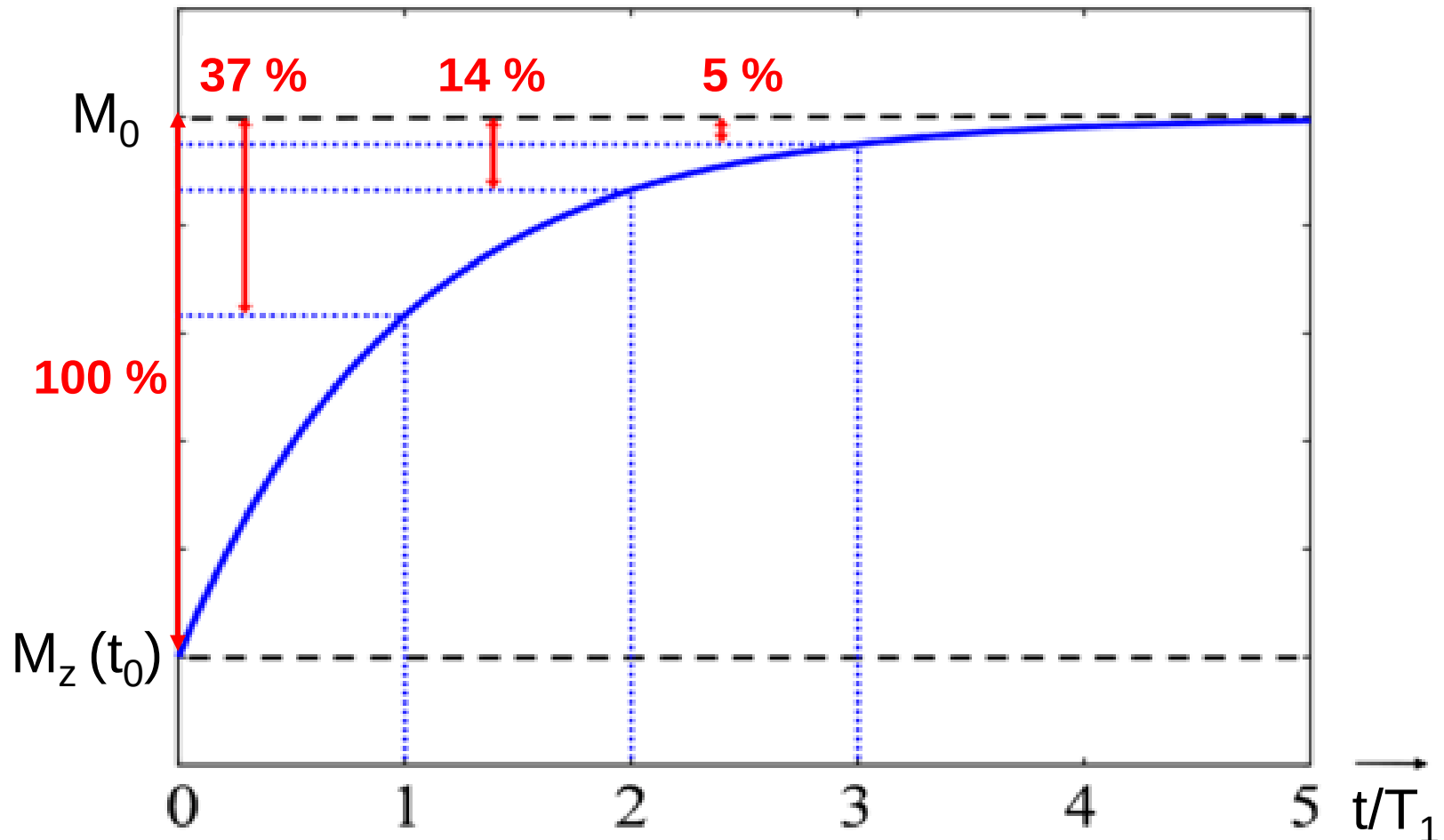


t/T_1

Relaxation T1,T2

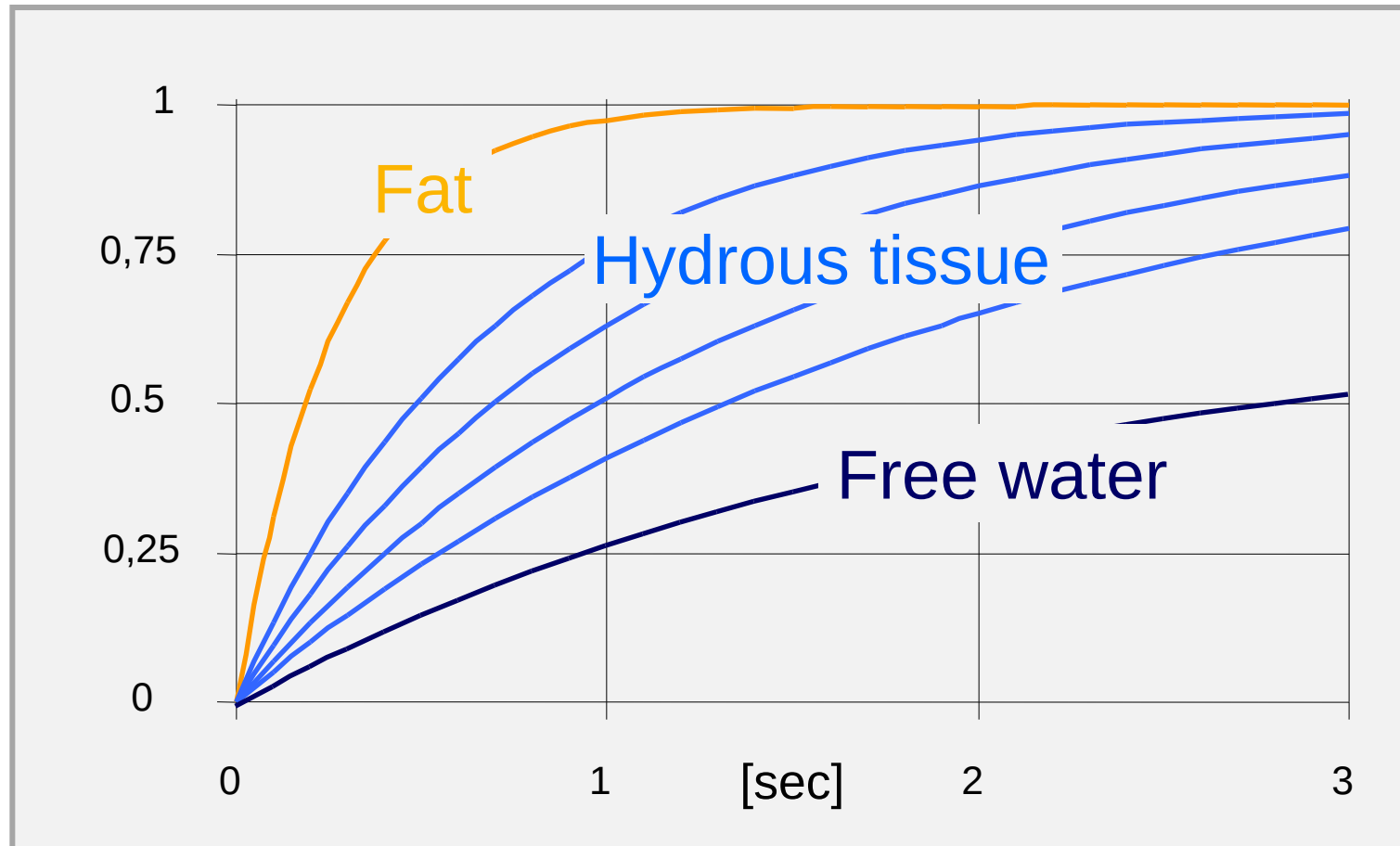
Exponential longitudinal relaxation with T1:

$$M_z(t) = M_0 - (M_0 - M_z(t_0)) \cdot \exp(-t/T_1)$$



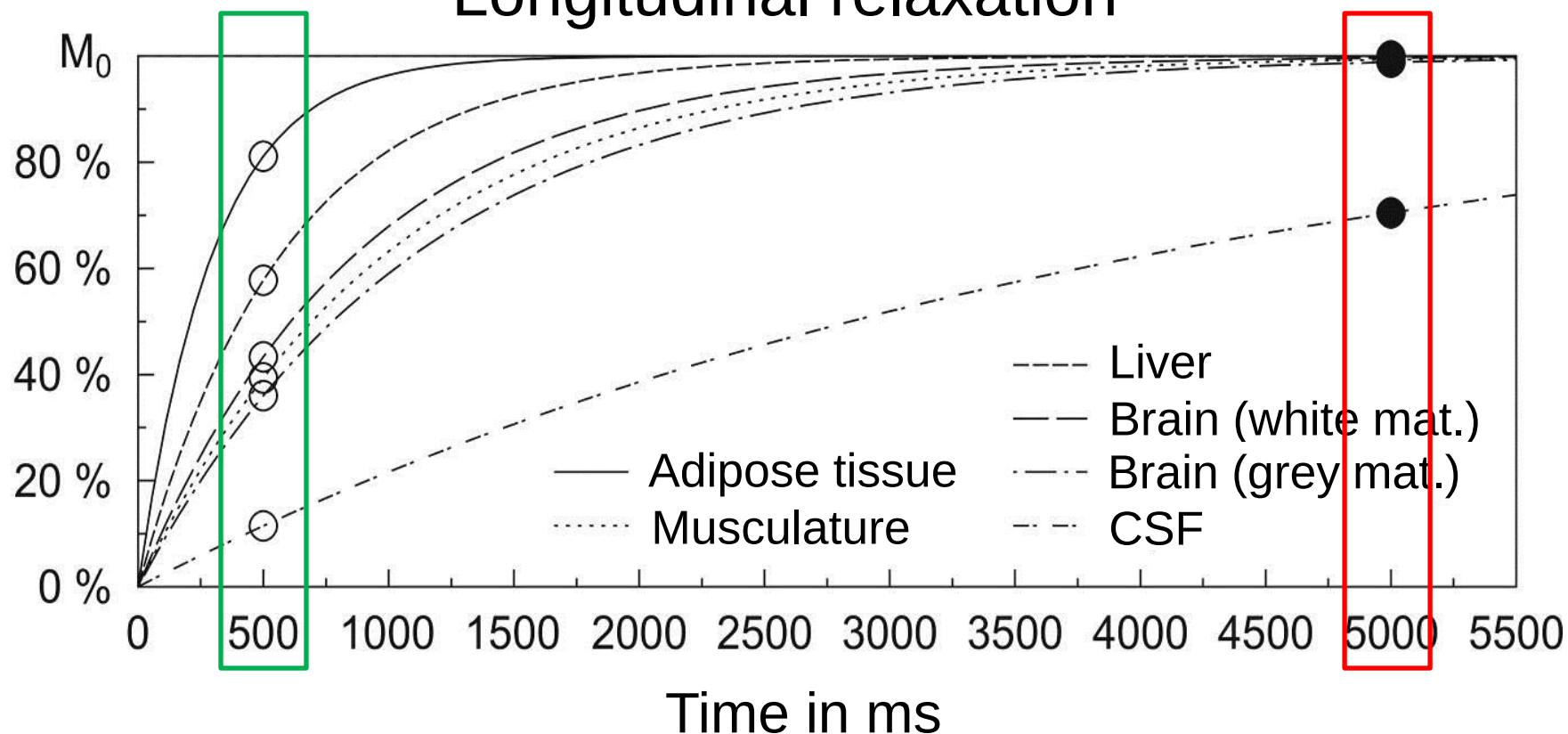
Relaxation T1,T2

Course of the longitudinal magnetization following a 90° excitation



Relaxation T1,T2

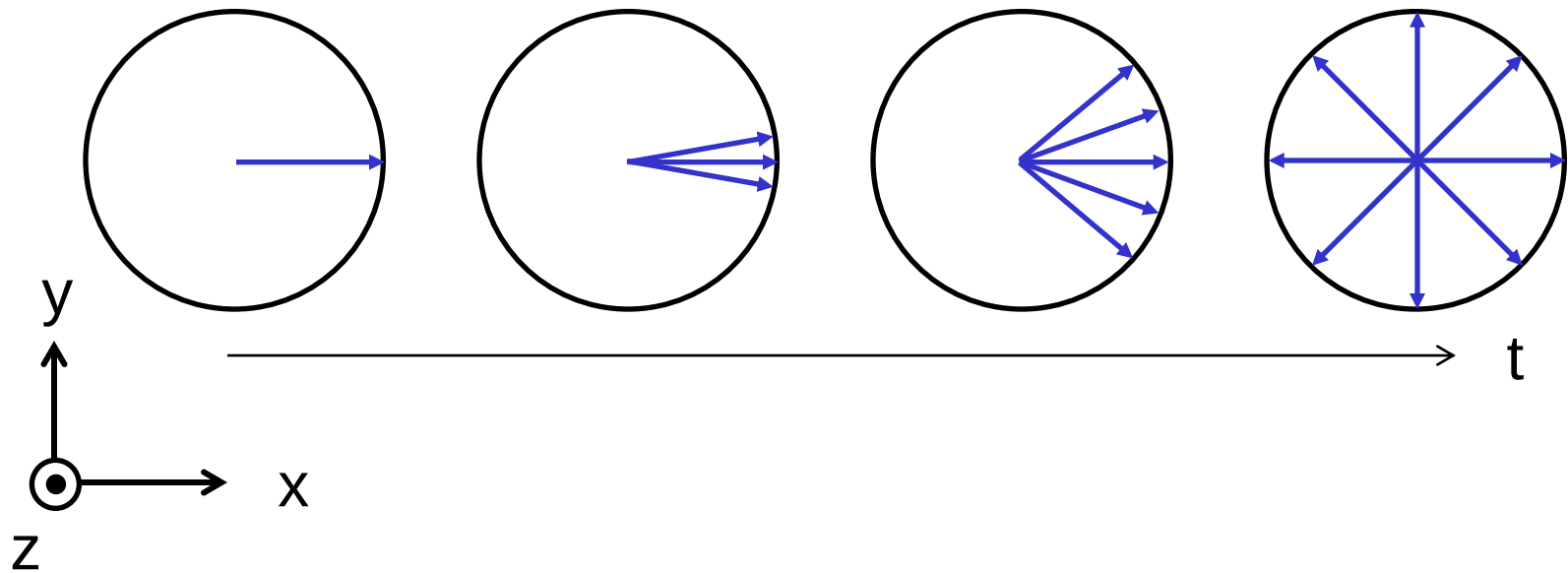
Longitudinal relaxation



high contrast

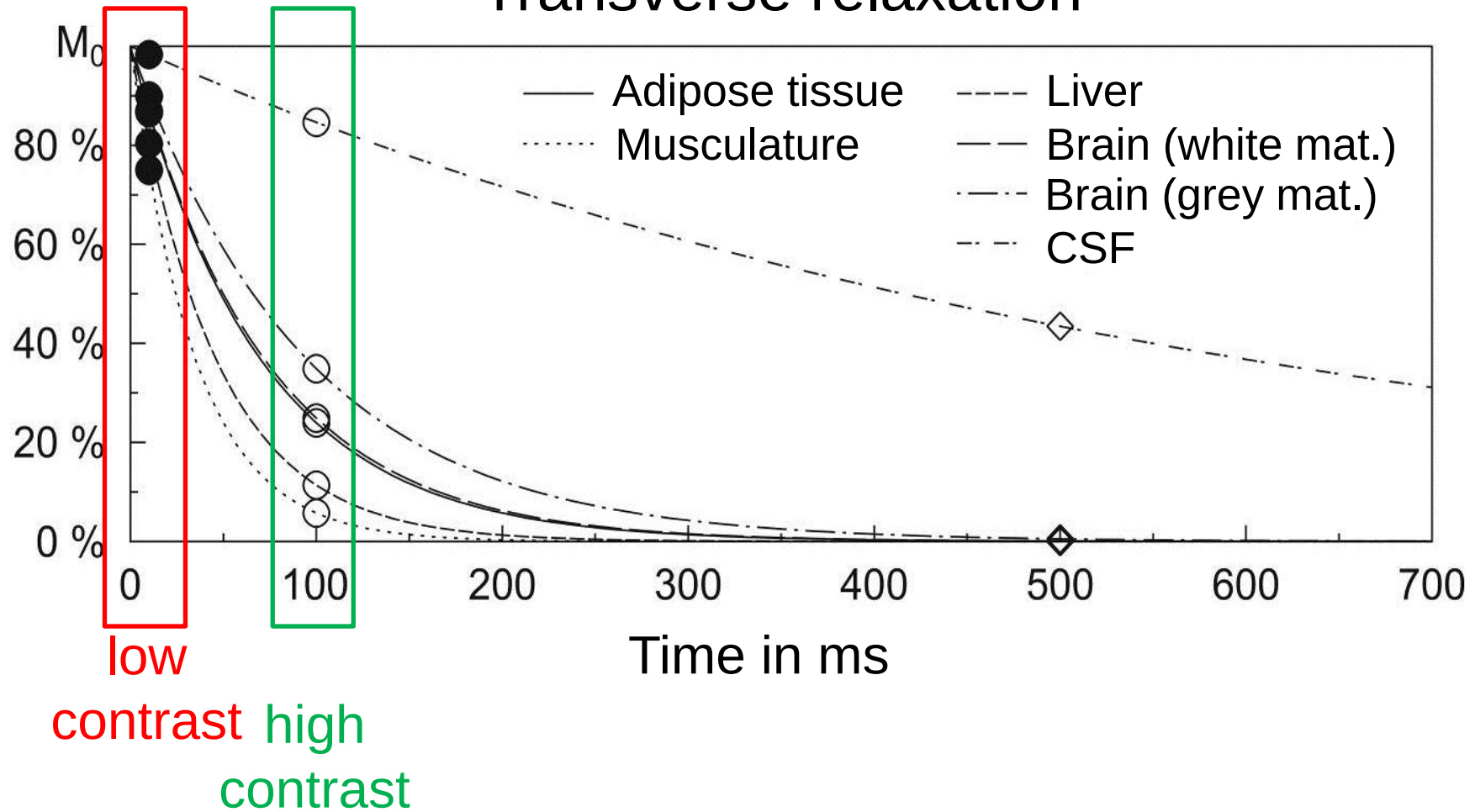
low/no
contrast

Relaxation T1,T2



Relaxation T1,T2

Transverse relaxation



Relaxation T1,T2

Relaxation times of tissues

Longitudinal relaxation time T1:

- Adipose tissue: **0,3 s**
(subcutaneous fat, belly fat, yellow bone marrow)
- Hydrous organ tissue: **≈ 1 s** (0,6 s to 1,2 s)
(e.g. liver, kidney, muscle, brain)
- Free aqueous liquid: **≈ 3 s**
(liquor, urine, some cysts)

Transversal relaxation time T2:

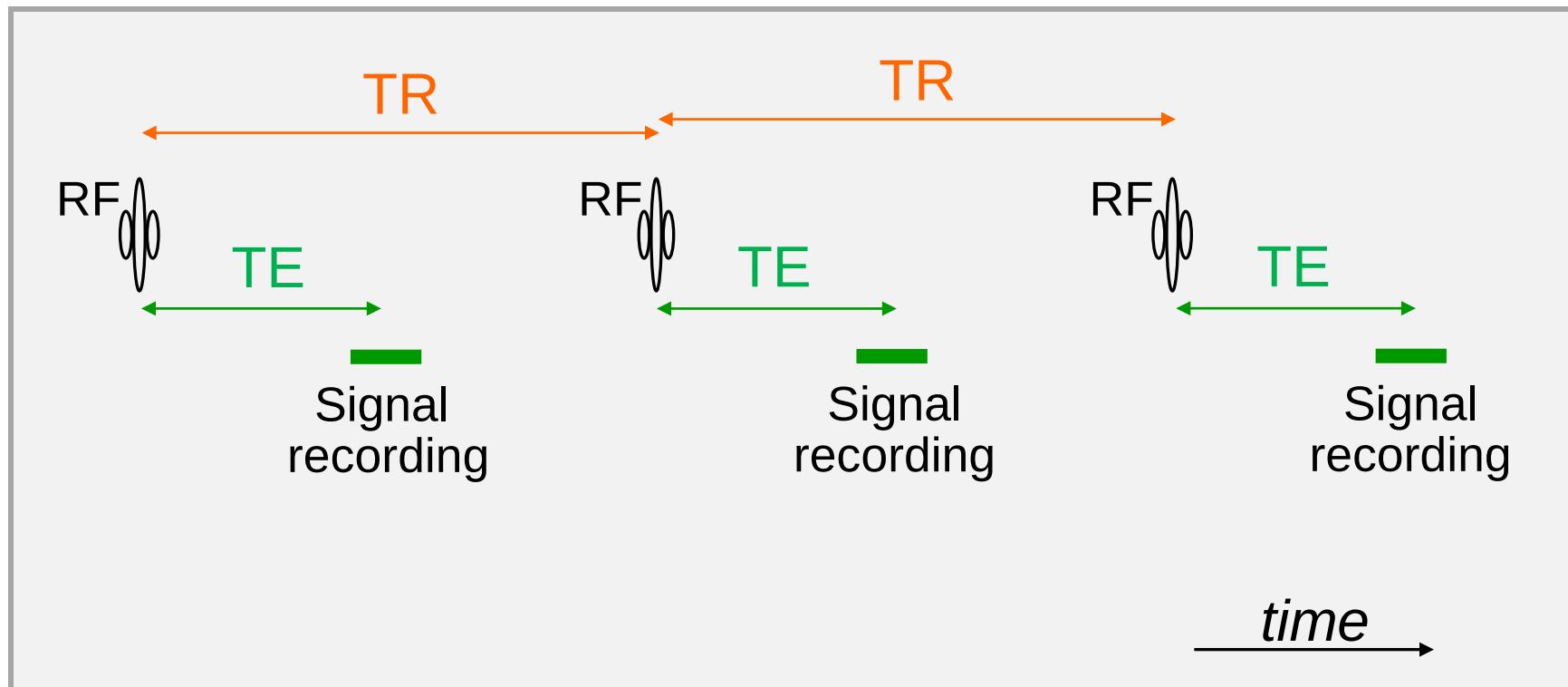
- Bones, tendons, bands, menisci **< 6 ms**
- Hydrous organ tissue: **30 ms to 100 ms**
(e.g. liver, kidney, muscle, brain)
- Free aqueous liquid : **> 1 s**
(liquor, urine, some cysts)

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Standard Contrasts

Process of RF-excitation and signal recording in MRI



Shorter repetition time **TR**: stronger T1-weighting
Longer echo time **TE**: stronger T2-weighting

Tissue Contrast

Standard-MRI

Proton density
weighting



long TR = 5 s
short TE = 10 ms

Relaxation weighting

T1-weighting



short TR = 0,5 s
short TE = 10 ms

T2-weighting

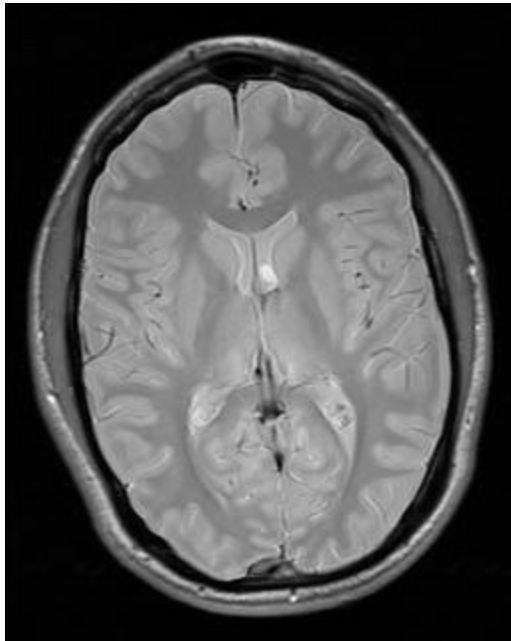


long TR = 5 s
long TE = 100 ms

Tissue Contrast

Standard-MRI

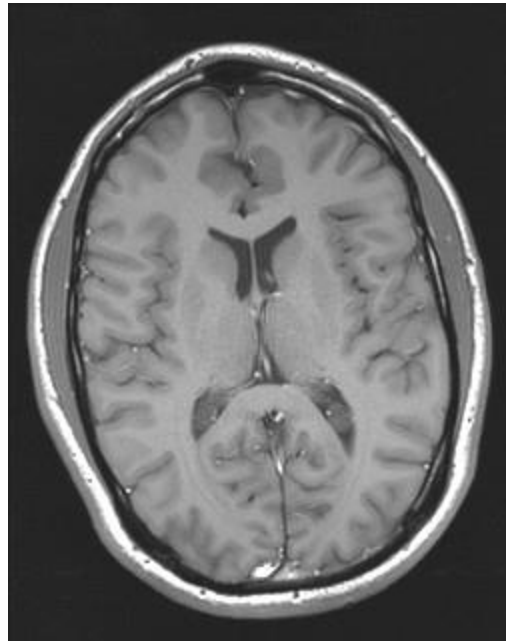
Proton density
weighting



long TR = 5 s
short TE = 10 ms

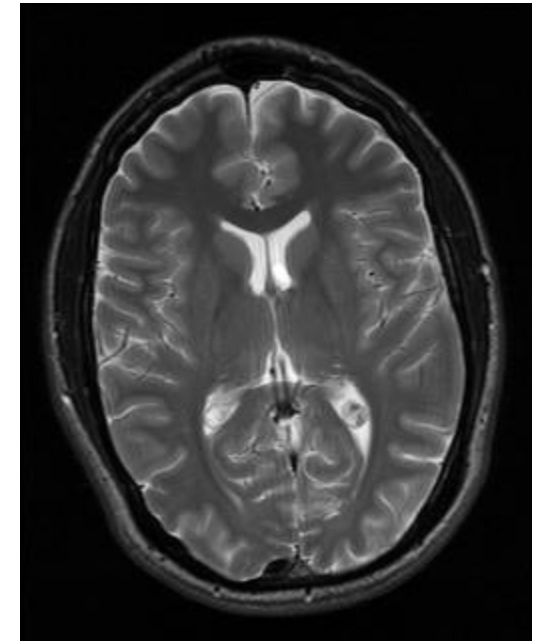
Relaxation weighting

T1-weighting



short TR = 0,5 s
short TE = 10 ms

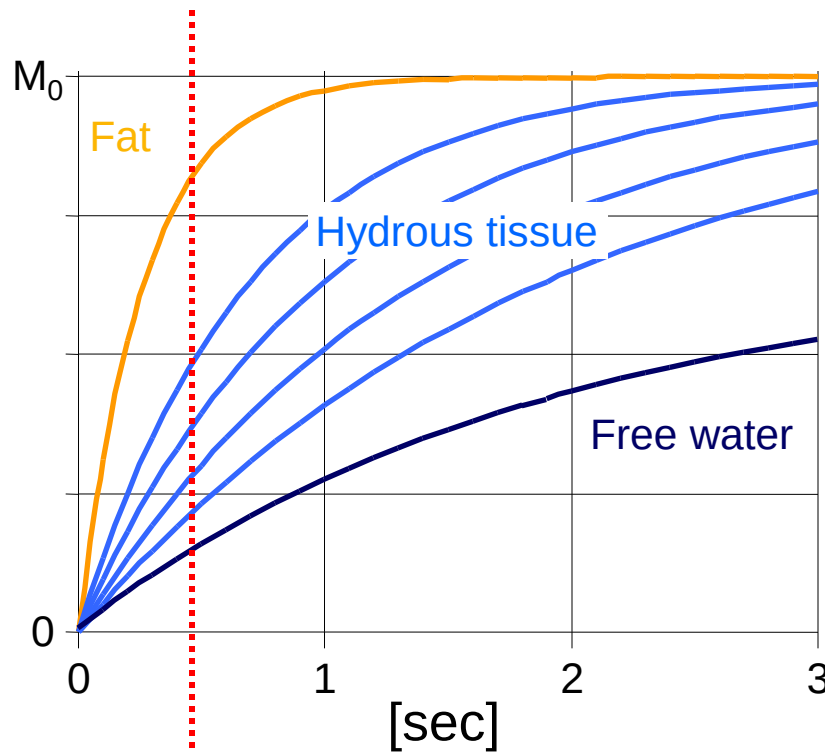
T2-weighting



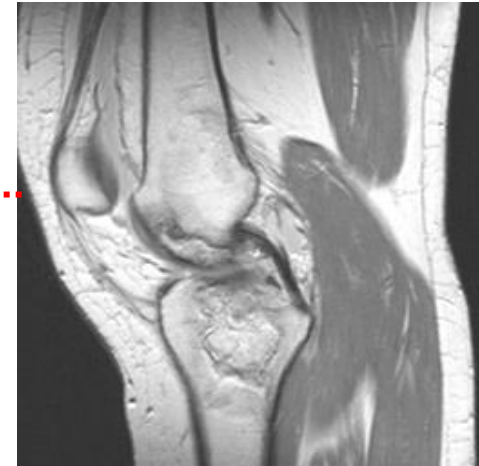
long TR = 5 s
long TE = 100 ms

Tissue Contrast

Time course of the longitudinal relaxation after RF-excitation



“Proton density”
weighting

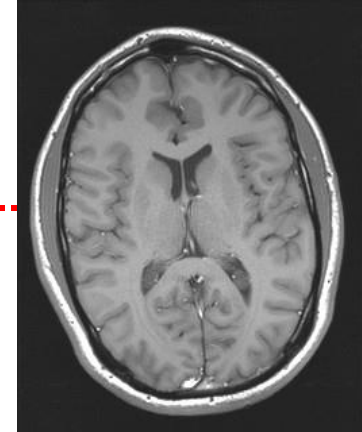


T1-weighting

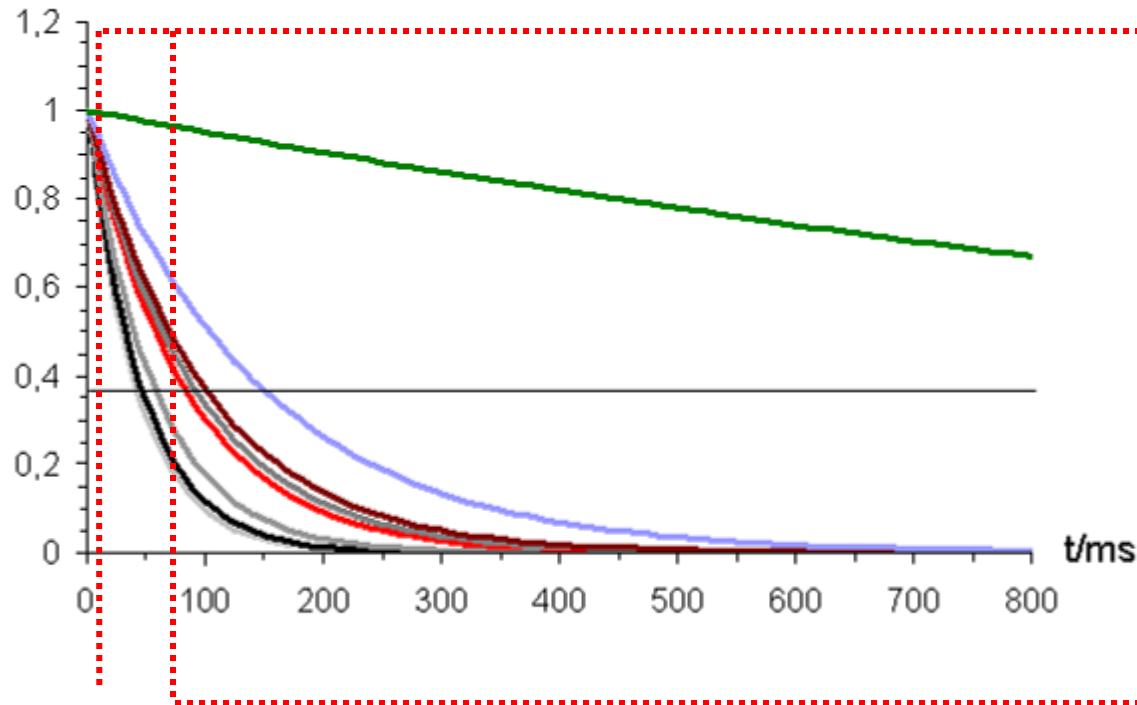
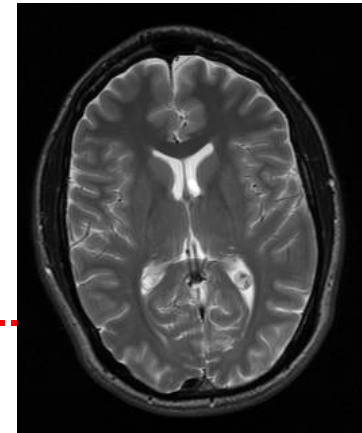


Tissue Contrast

T1-weighting



T2-weighting



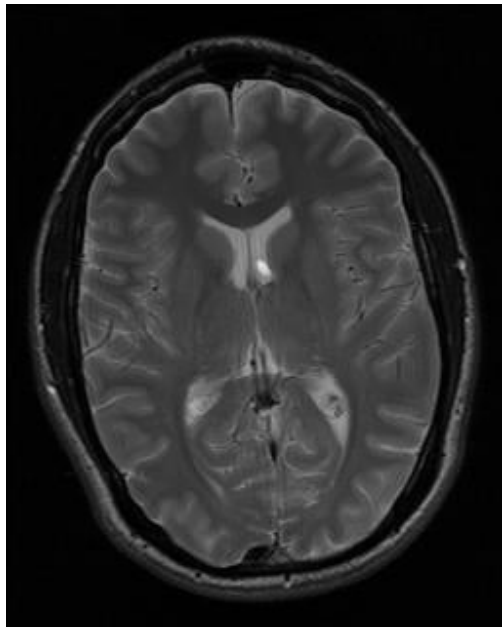
Tissue Contrast

Setting of TR and TE for different contrast weightings (valid for SE and TSE sequences)

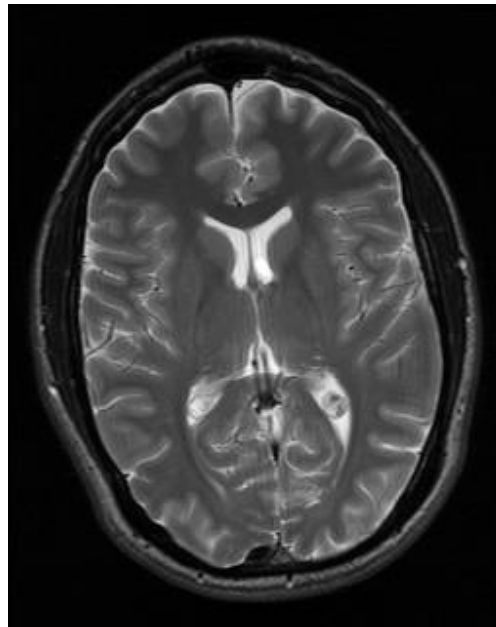
- Proton density weighting
 - TR long ($> 4\text{ s}$)*
 - TE short ($< 12\text{ ms}$)*
- Relaxation weighting (T1, T2)
 - T1-weighting:
 - $300\text{ ms} < TR < 800\text{ ms}$;*
 - $TE < 12\text{ ms}$*
 - T2-weighting:
 - $TR > 4\text{ s}$; $TE > 80\text{ ms}$*

Tissue Contrast

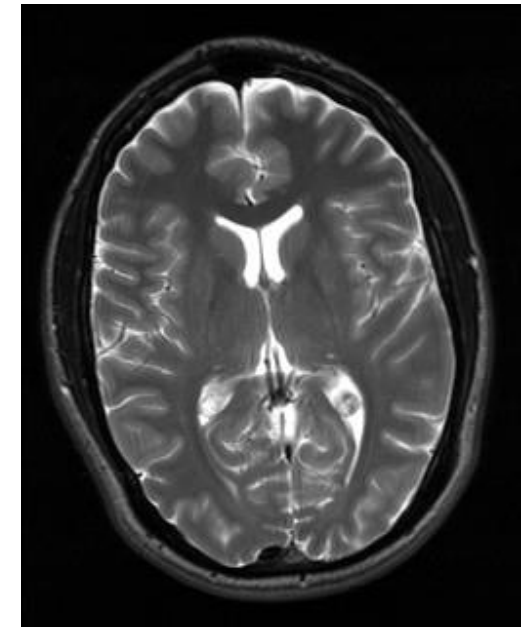
- For pure T2-weighting TR should be (in theory) infinitely long in order to avoid T1 effects.
- For TR = 5 s all tissues excluding free water are largely “relaxed”.
- Free water gains even more signal intensity if TR gets extended over 5 s.



TR = 2,5 s
TE = 100 ms



TR = 5 s
TE = 100 ms



TR = 10 s
TE = 100 ms

Tissue Contrast

A prolongation of TE to 400 ms or longer leads to a selective depiction of areas containing aqueous fluids.

Applications: MR myelography, MR ventriculography, MR urography, MRCP (cholangio and pancreaticography)



T2-weighted
data set
TE= 500 ms!

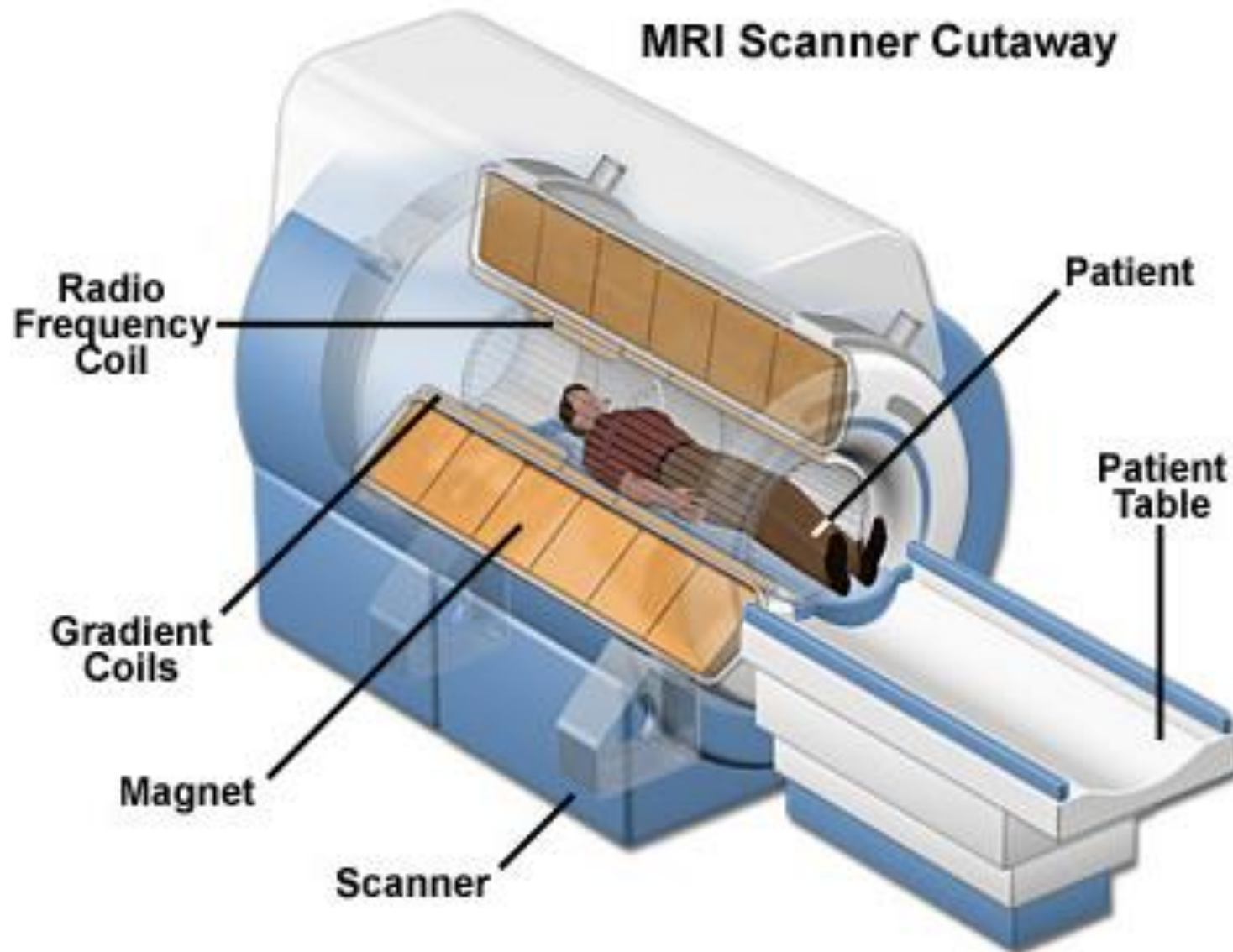
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Hardware Components

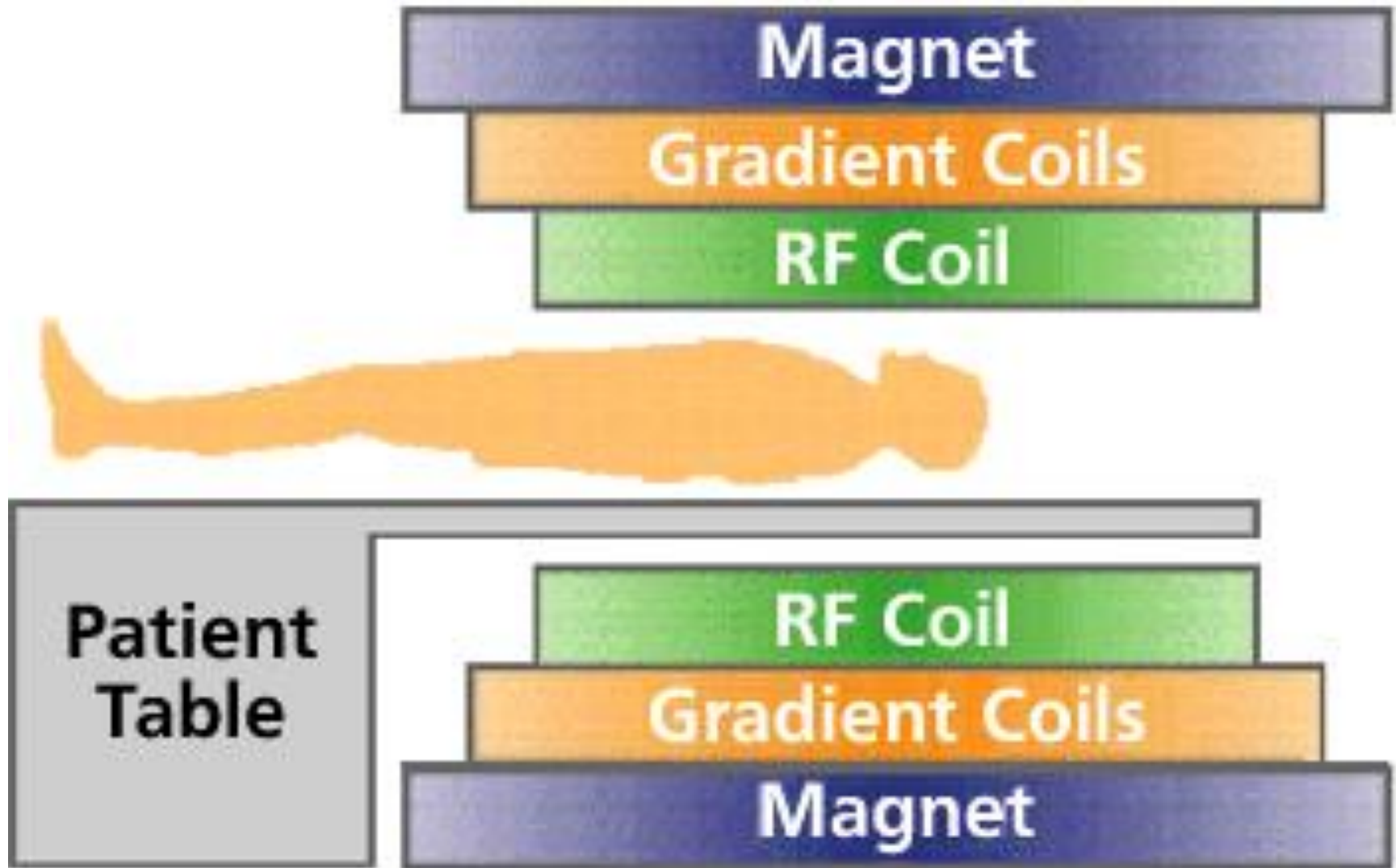
- Magnet
 - generates a homogeneous static magnetic field B_0
 - field strength 0.2 to 3.0 (7.0) Tesla for clinical systems
- Gradient system (+ shim coils)
 - rapidly switchable additional spatially inhomogeneous magnetic fields for slice selection and spatial encoding of signals. Respective fields are superimposed on the static field.
- RF transmitter
 - RF-transmitter for generation of a spatially homogeneous RF magnetic field for spin excitation operating at the Larmor frequency of ^1H nuclei
 - 8.5 MHz (0.2 T) to 128 MHz (3.0 T) with power up to 30 kW !!
- RF receiver (mostly movable coils)
 - highly sensitive RF resonance circuit with a geometric configuration adapted to the body part of interest

Hardware Components



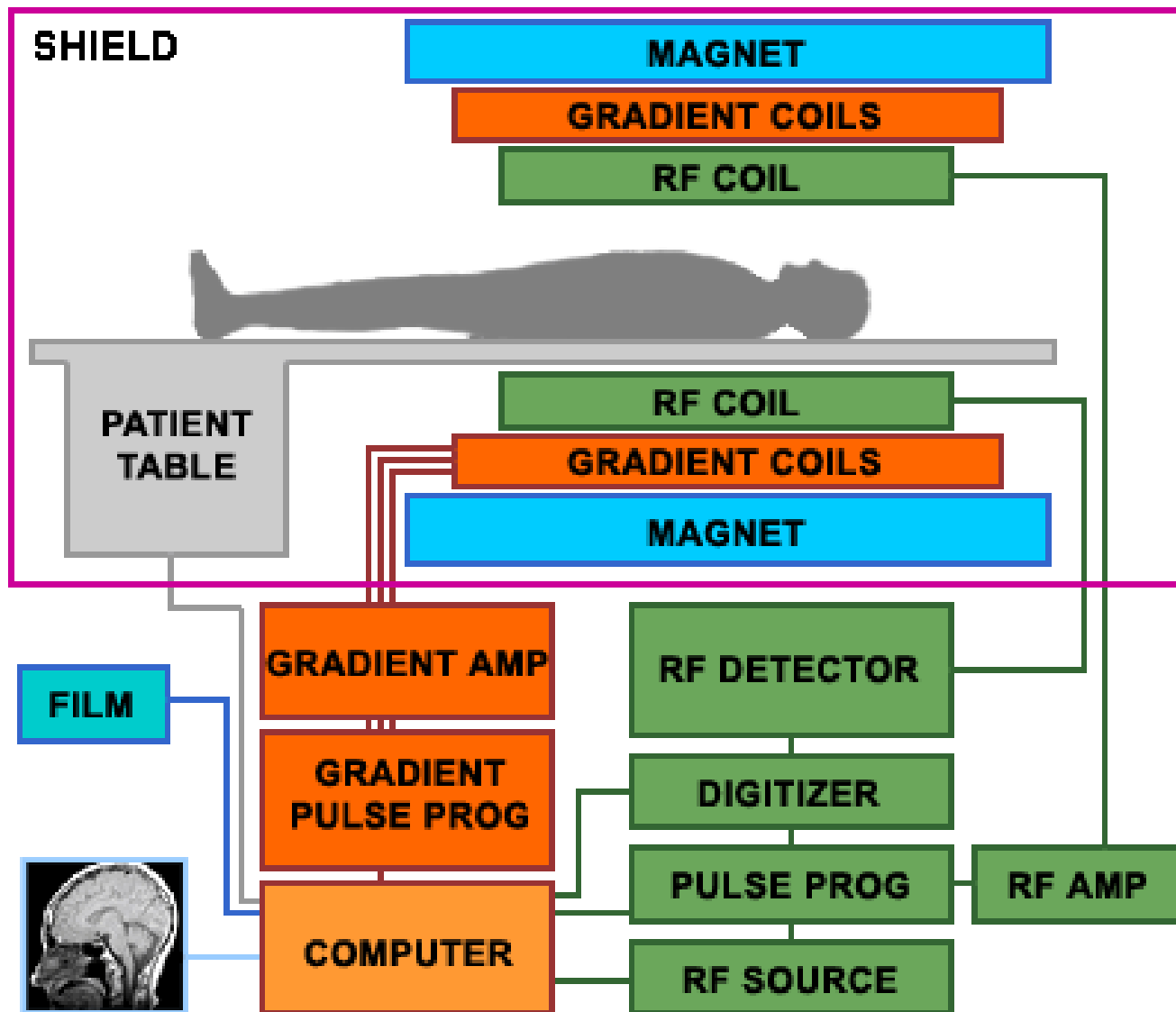
Hardware Components

Specific components of an MRI system



Hardware Components

MRI system and peripheral components



Magnets

Magnet types

Resistive electromagnet

- mostly water-cooled
- field strength up to 0.3 Tesla

Permanent magnet

- ferrite, samarium-cobalt or neodymium magnets
- of very heavy weight
- field strength up to 0.35 Tesla

Superconducting magnet

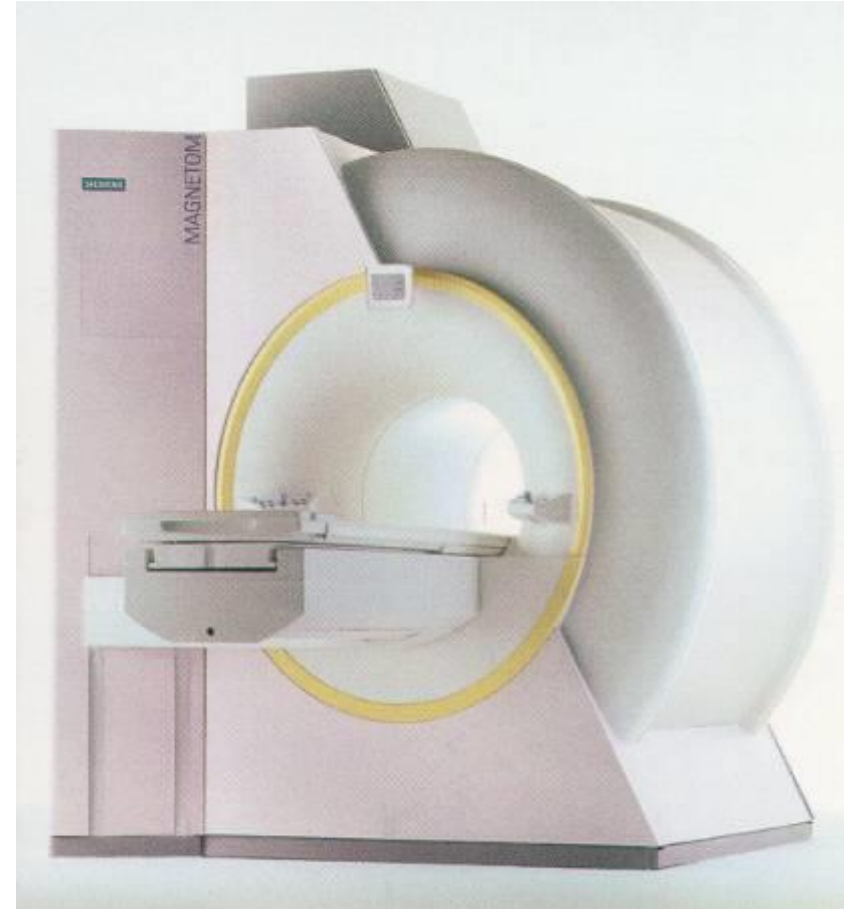
- coil windings of niobium-titanium filaments (critical temp. 10 K)
- liquid helium cooling (boiling temp. 4,2 K)
- field strength currently up to 9.4 T for whole-body systems
- used for most clinical MR units, field strength is mostly 1.5 T

Magnets

Different designs and types of magnets are used in the clinic:



0.2 Tesla, open,
resistive magnet (6 tons),
or permanent magnet (11 tons)



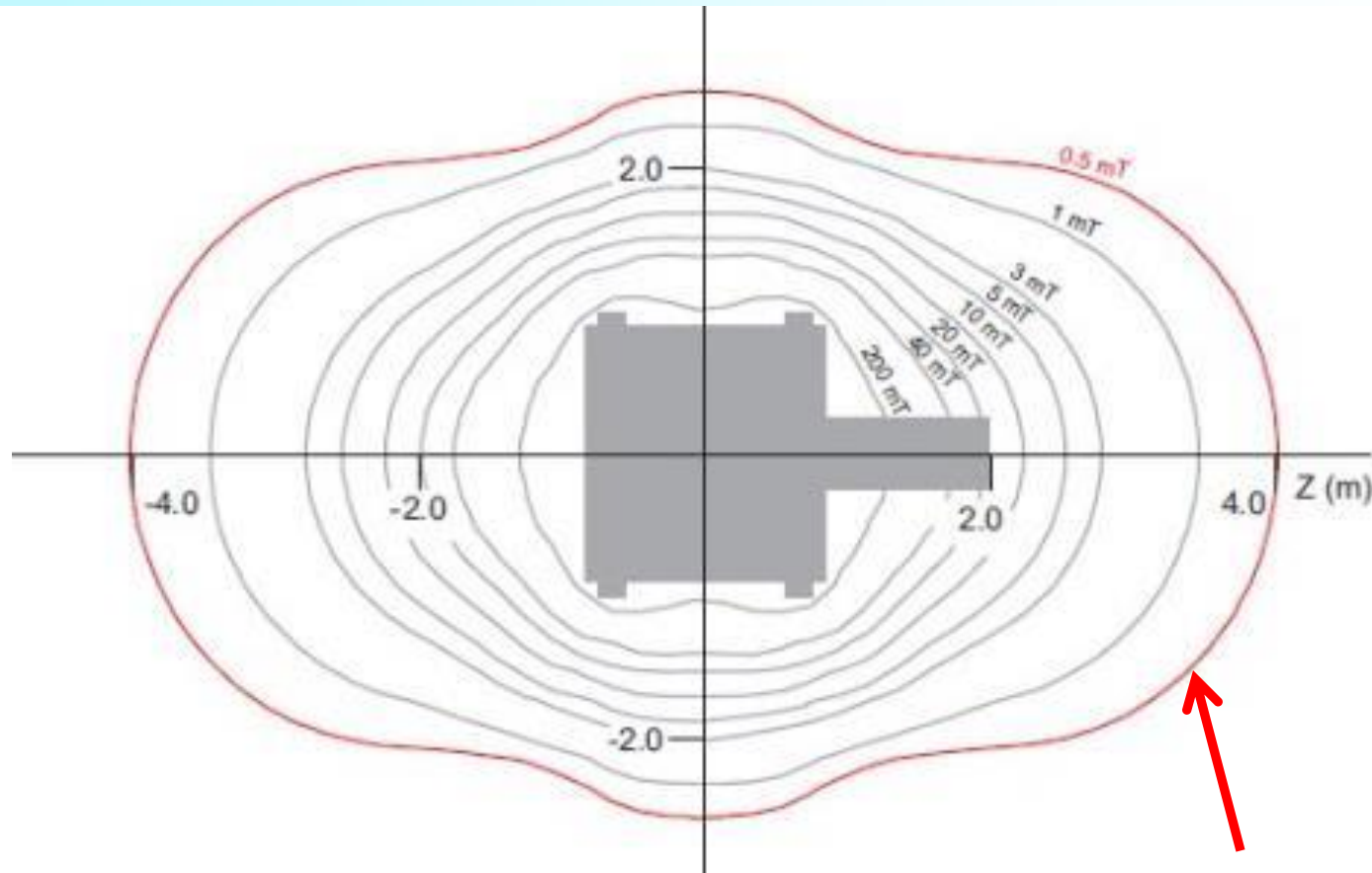
1.5 Tesla, closed bore
superconducting magnet

Magnets

Superconducting magnet of a 3T clinical MRI system



Fringe Field of a 3 Tesla Whole Body System



Earth magnetic field:
 $0.00005 \text{ T} = 0.05 \text{ mT}$
 $= 50 \mu\text{T} = 0.5 \text{ Gauss}$

5 Gauss (= 0.5 mT)
is indicated by a
red line on the floor
Critical items must stay outside!

Forces

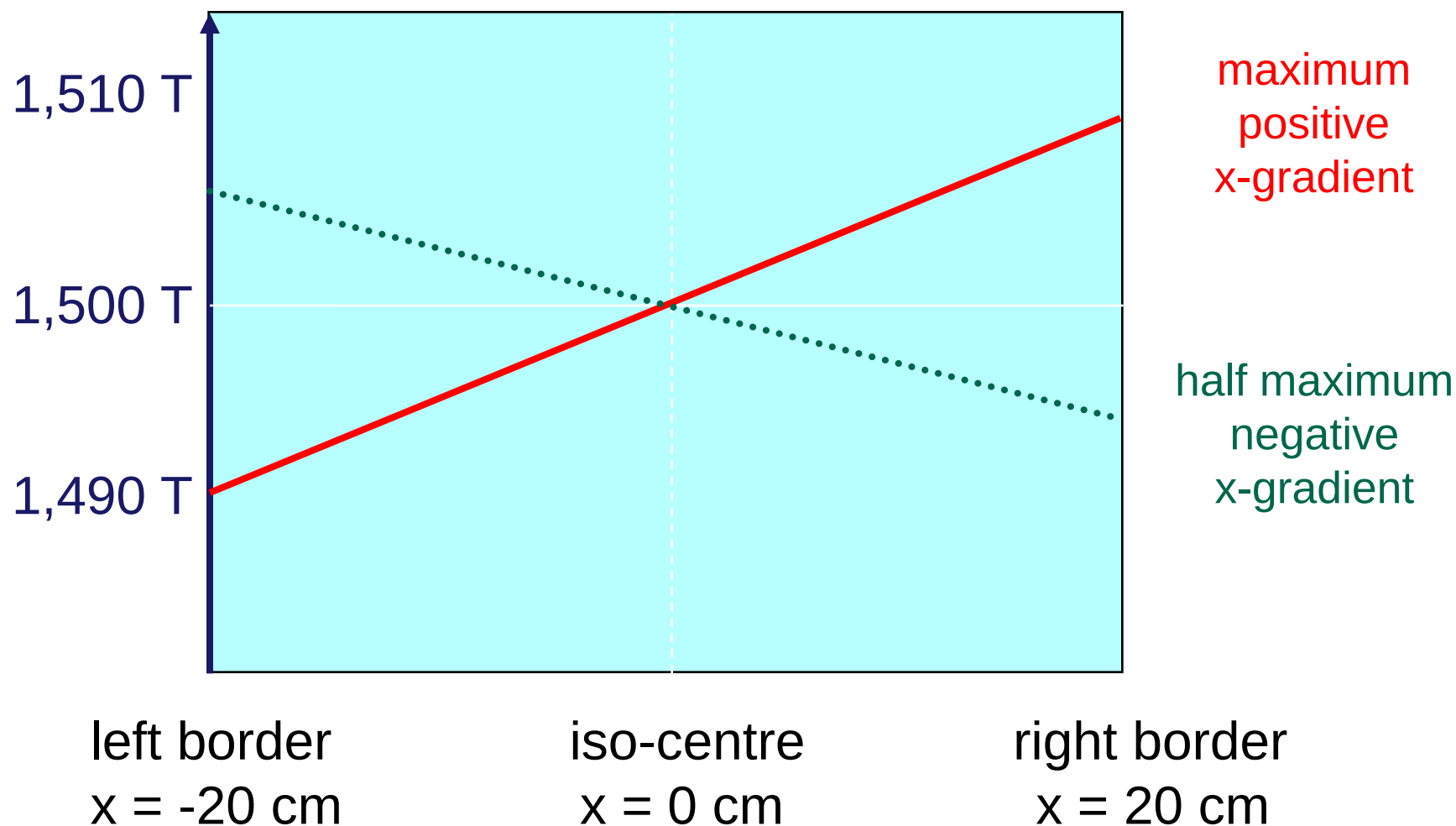
ely strong!



Gradient System

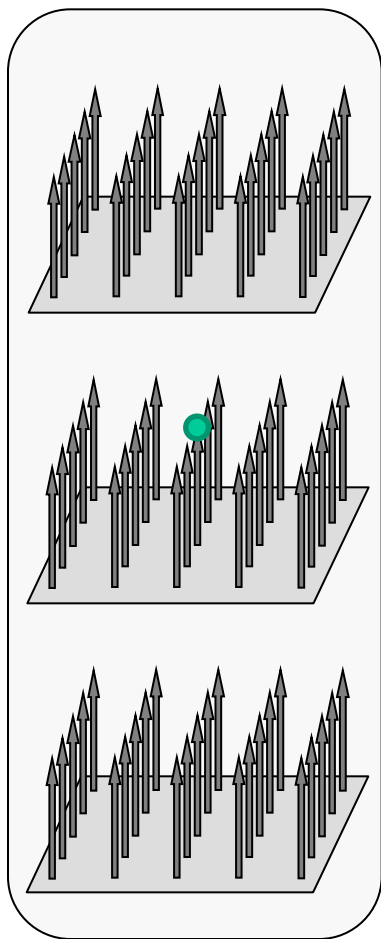
Effects of gradient switching along the x-axis

magnetic field strength

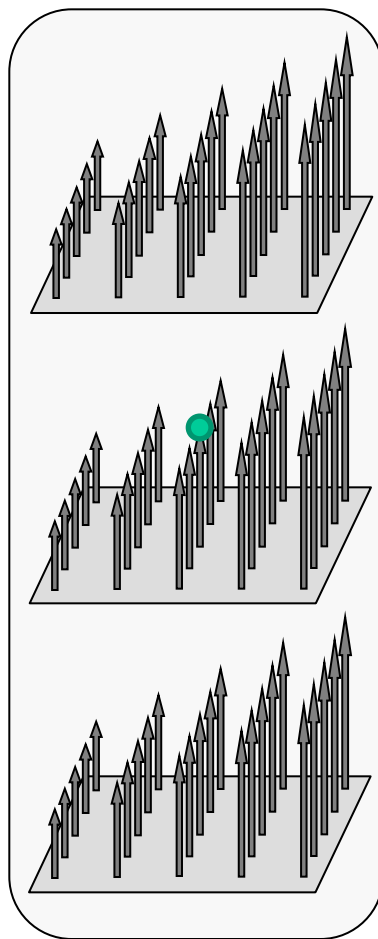


Gradient System

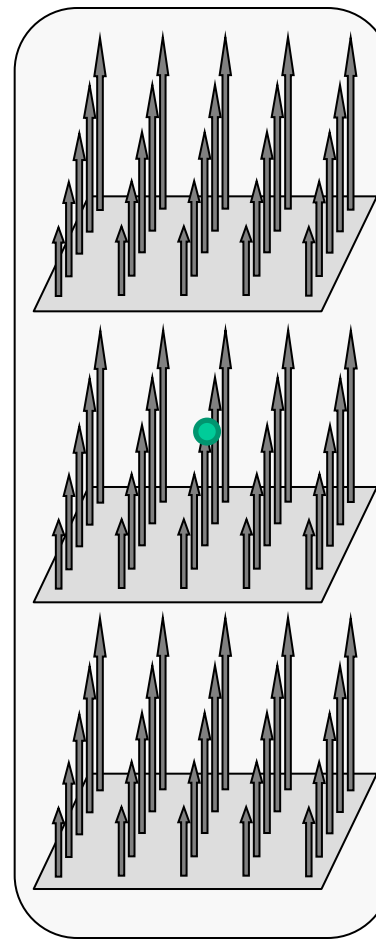
no field gradient



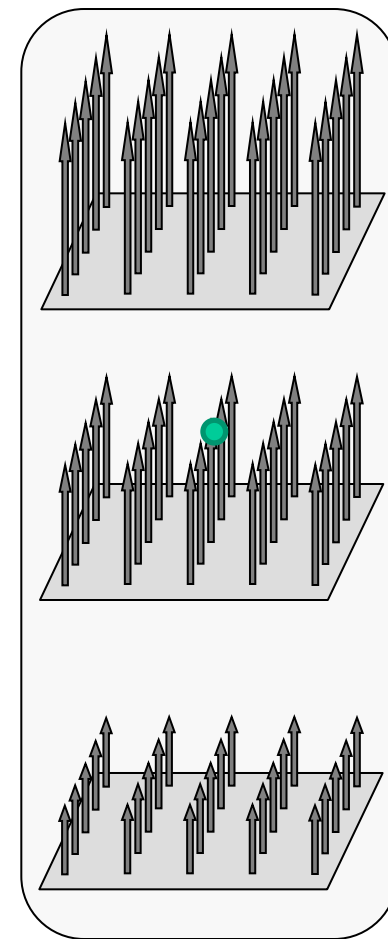
x - gradient



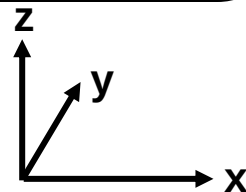
y - gradient



z - gradient



$\uparrow B_0$ (static magnetic field)

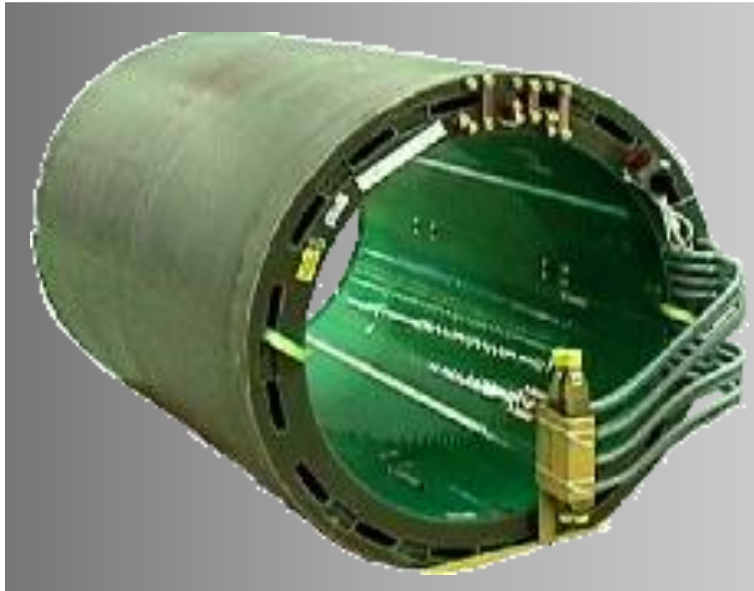


● iso-center: field strength remains constant despite gradient switching

Gradient System

Gradient coils of a clinical whole body system

- Gradient coils provide constant field gradients along x-, y- or z- direction (z-direction is along B_0)
- An MRI system works with three independent gradient coil systems
- Typical rise times from zero to maximum gradient strength amount to approx. 0.2 ms
- Maximum gradient amplitudes of whole body systems are 30-80 mT/m



RF Transmitter Coils

Aims

- homogeneous RF excitation over a large volume
- restriction of RF energy absorbed by the body or body parts (especially at higher field strength)

Types of RF transmitter coils

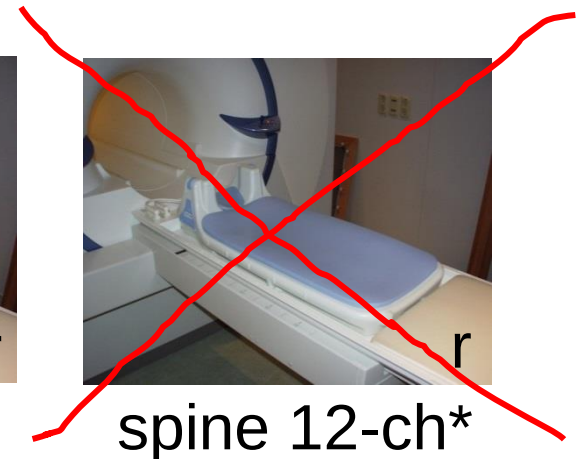
- by the **large transmitter coil** inside the casing of the MR unit,
or
- by **mobile volume coils**, serving as transmitter and receiver



„body coil“



knee 1-ch



spine 12-ch*

RF Receiver Coils



wrist/hand 1-ch



knee 1-ch



knee 8-ch



spine 12-ch



torso/cardiac 8-ch



head 1-ch



head 1-ch



head 8-ch

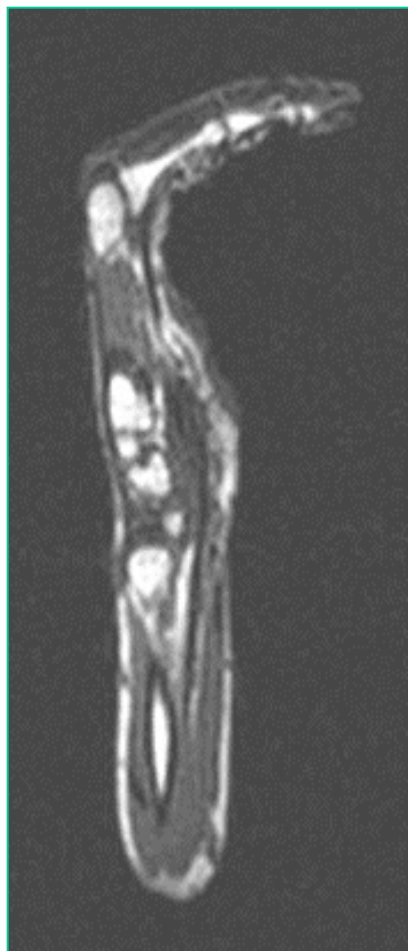
RF Receiver Coils

Aims

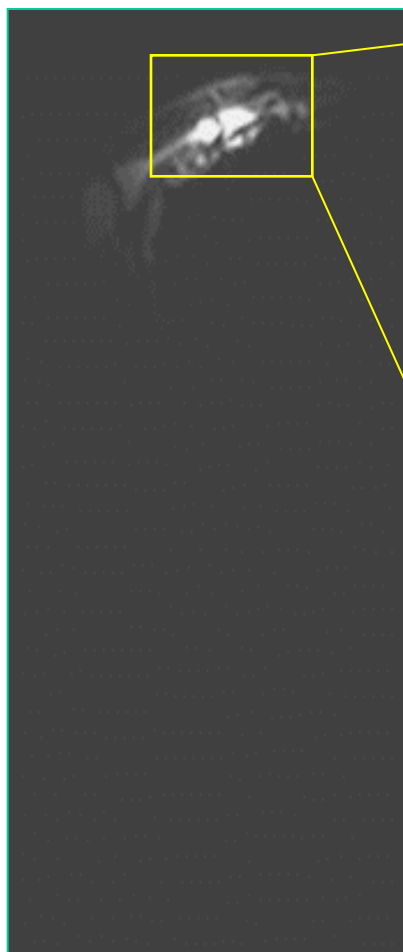
- high sensitivity to low intensity MR signals
 - low noise level and low sensitivity to electromagnetic sources in the surroundings (thermal noise)
 - homogeneous level of sensitivity over (extended) body regions under investigation
- Suitable compromises are strongly dependent on the body part under investigation
- Many different receiver coil systems are available for different types of examinations

RF Receiver Coils

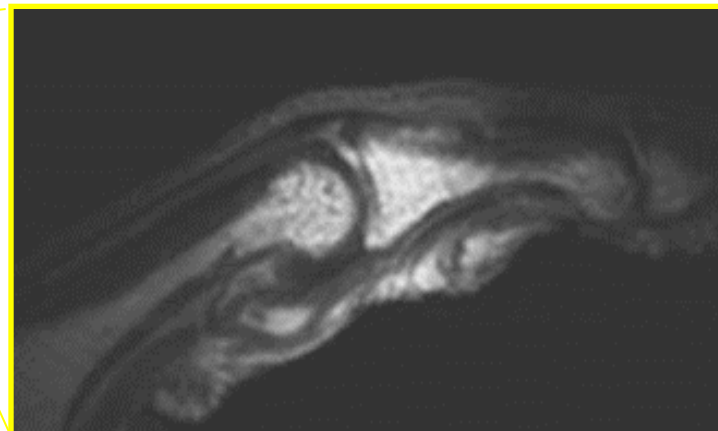
Comparison of large and small single coils



body coil



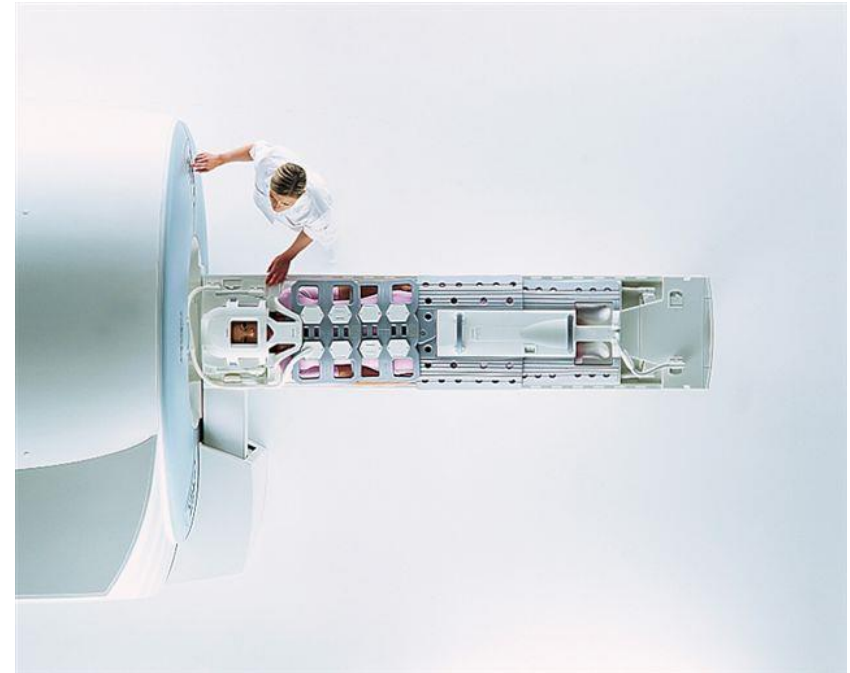
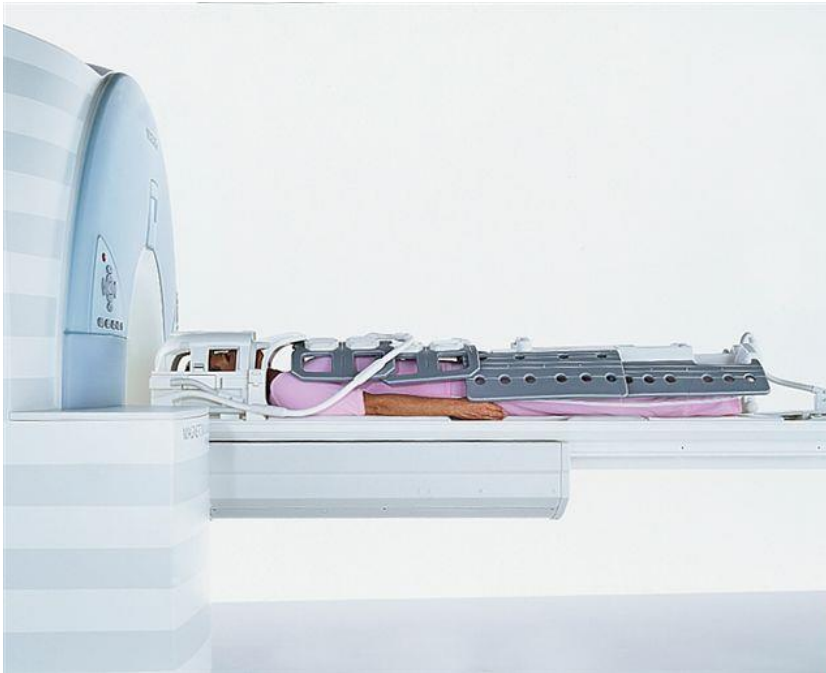
finger coil



finger coil provides higher spatial resolution, but locally restricted sensitivity

RF Receiver Coils

Whole body coverage using arrays of receiver coils



Coverage of the entire body by approx. 70-120 receiver coils allowing whole-body examinations without repositioning of the patient by a technician

Whole Body MRI



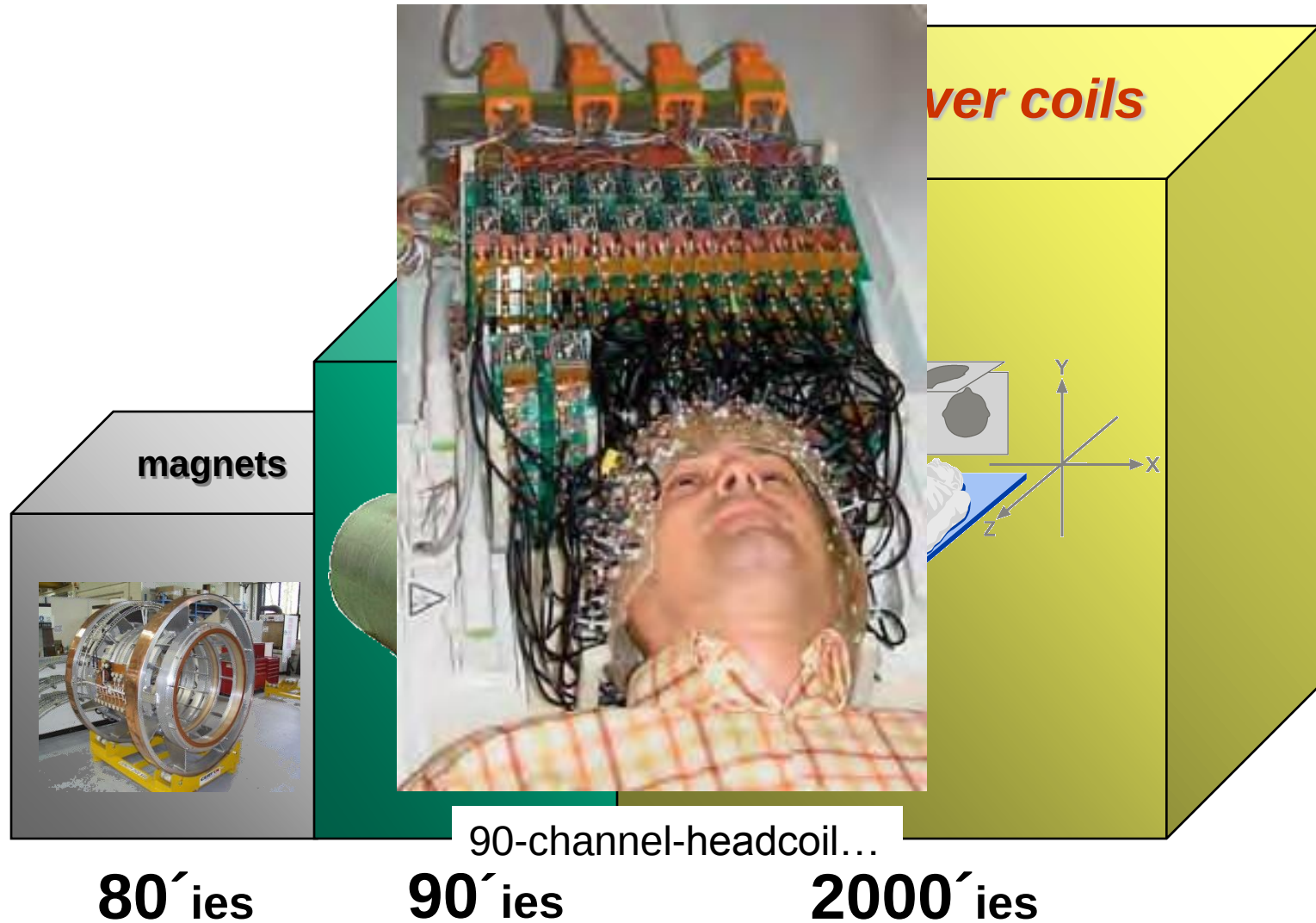
T1-weighted
spin echo

MR-angiography:
cardiovascular
system



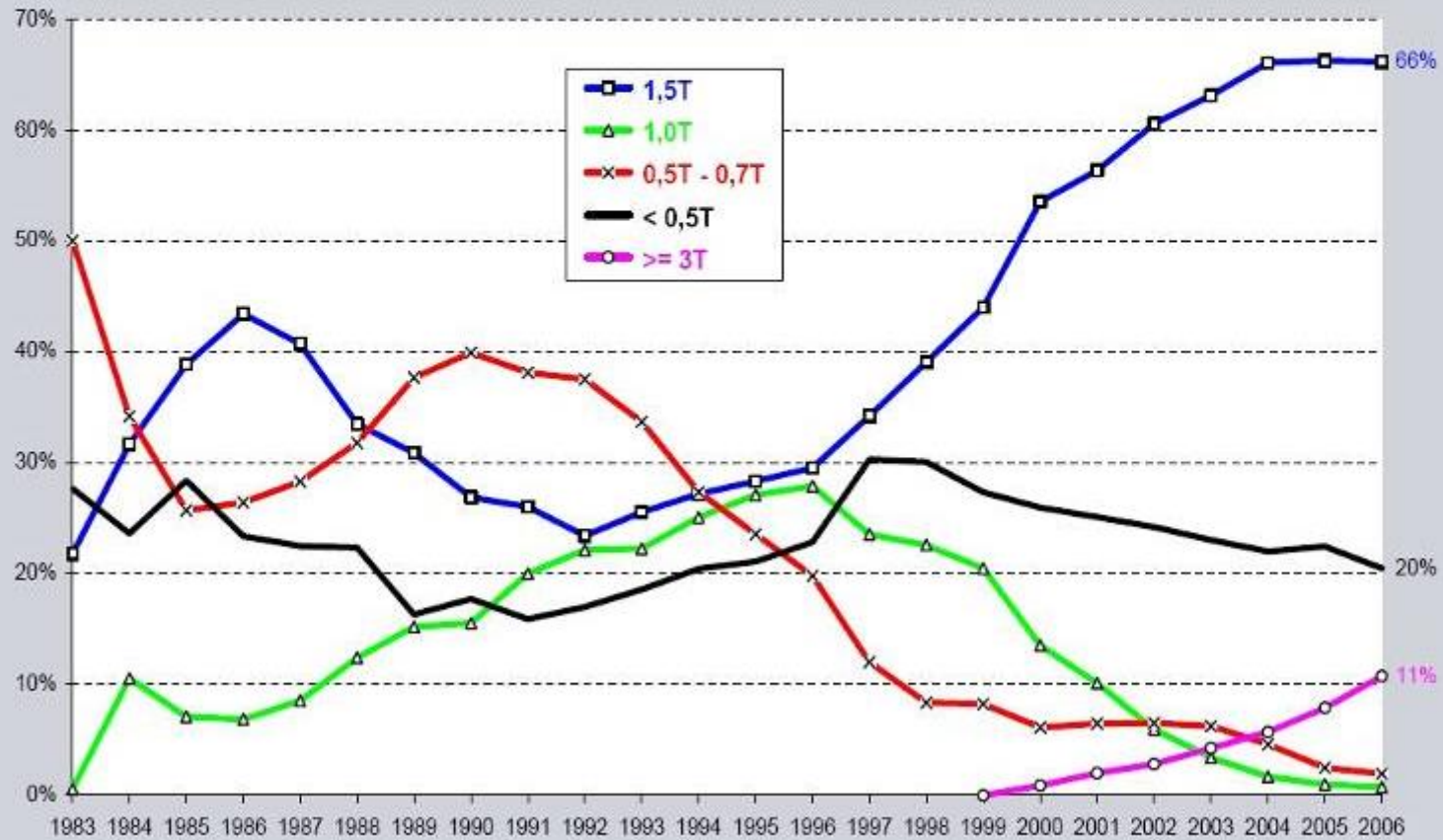
Steps in development

major interest in...



Steps in development

MR Produkte weltweit

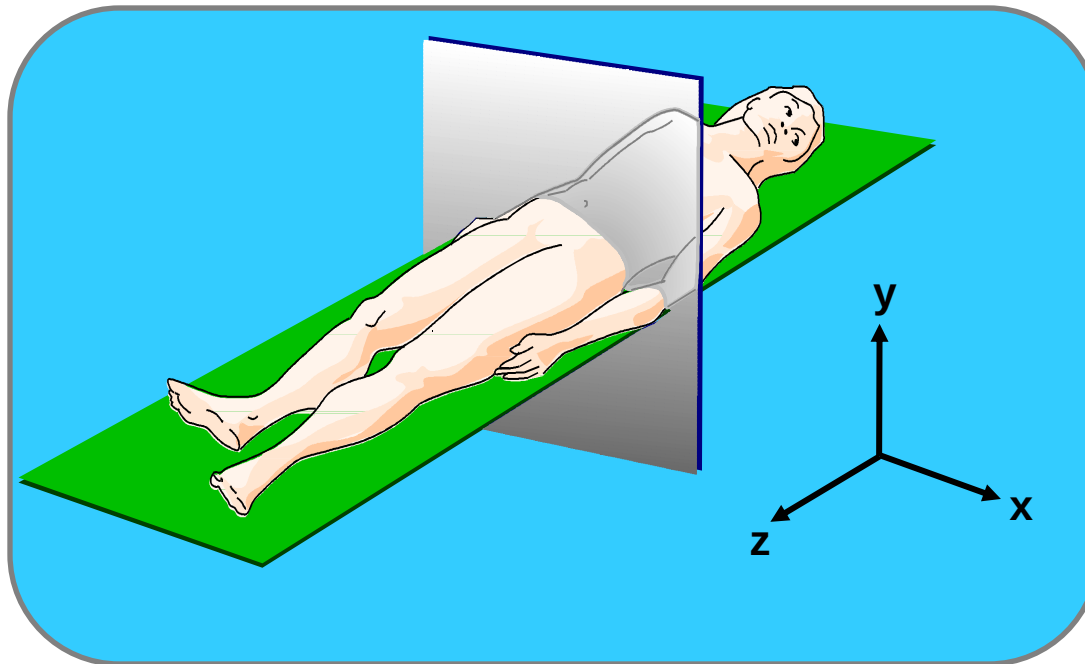


Outline

- **Basic principles of magnetic resonance:**
spins, magnetization, RF excitation, relaxation
- **Standard contrasts:**
proton density, T1 and T2 weighting
- **Hardware components of an MR imaging system:**
static magnetic field, gradient fields, RF field T/R
- **Cartesian spatial encoding:**
2D and 3D encoding
- **Standard sequences:**
gradient echo, spin echo, and fast spin echo

Principle of Imaging

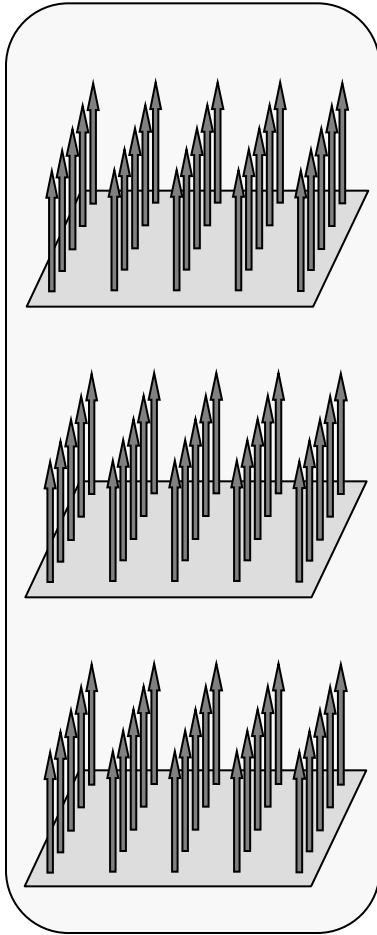
Slice recording in 3-dimensional space



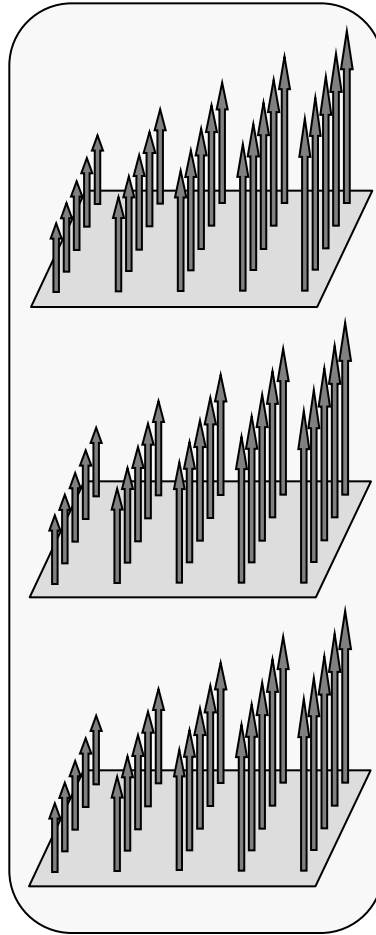
1. Slice selective excitation (z)
2. Frequency encoding of signals (x)
3. Phase encoding of signals (y)

Principle of Imaging

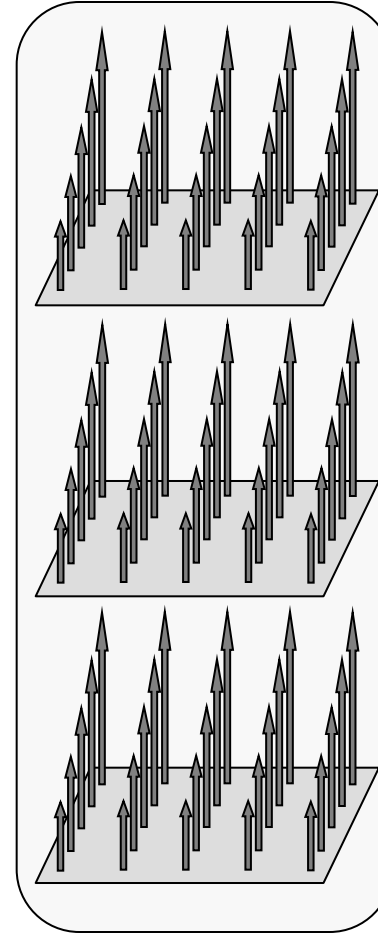
No field gradient



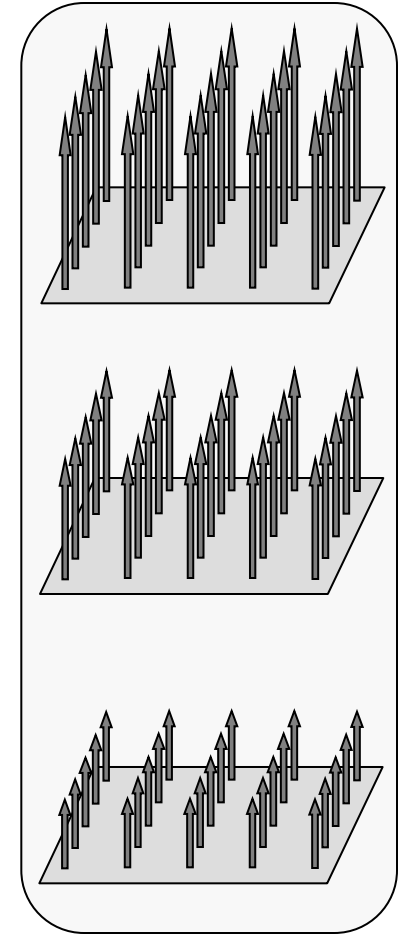
x - gradient



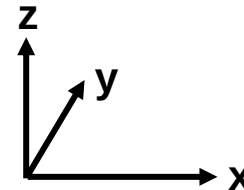
y - gradient



z - gradient

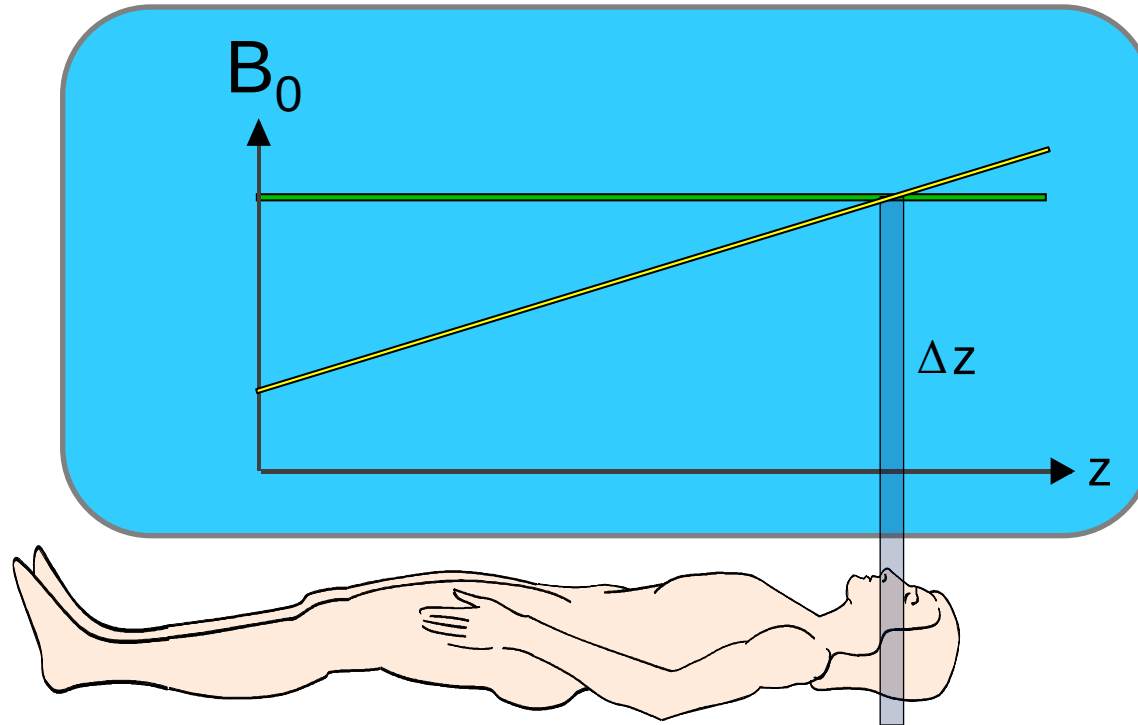


 B_0 (static magnetic field)



Principle of Imaging

Slice selective excitation



Simultaneous actions:

- Switching a gradient in z-direction (slice selection gradient)
- Application of a frequency band selective RF pulse

excitation/inversion of magnetization

Aim:

- Excitation, inversion of magnetization within a few ms
- magnetization in a defined (rectangular) frequency band has to experience the desired flip angle
- Different pulse forms (Gauß, sinc, Hermite, BURP) – optimization is/was an important feature some decades ago
- adiabatic pulses: best profile but high SAR!

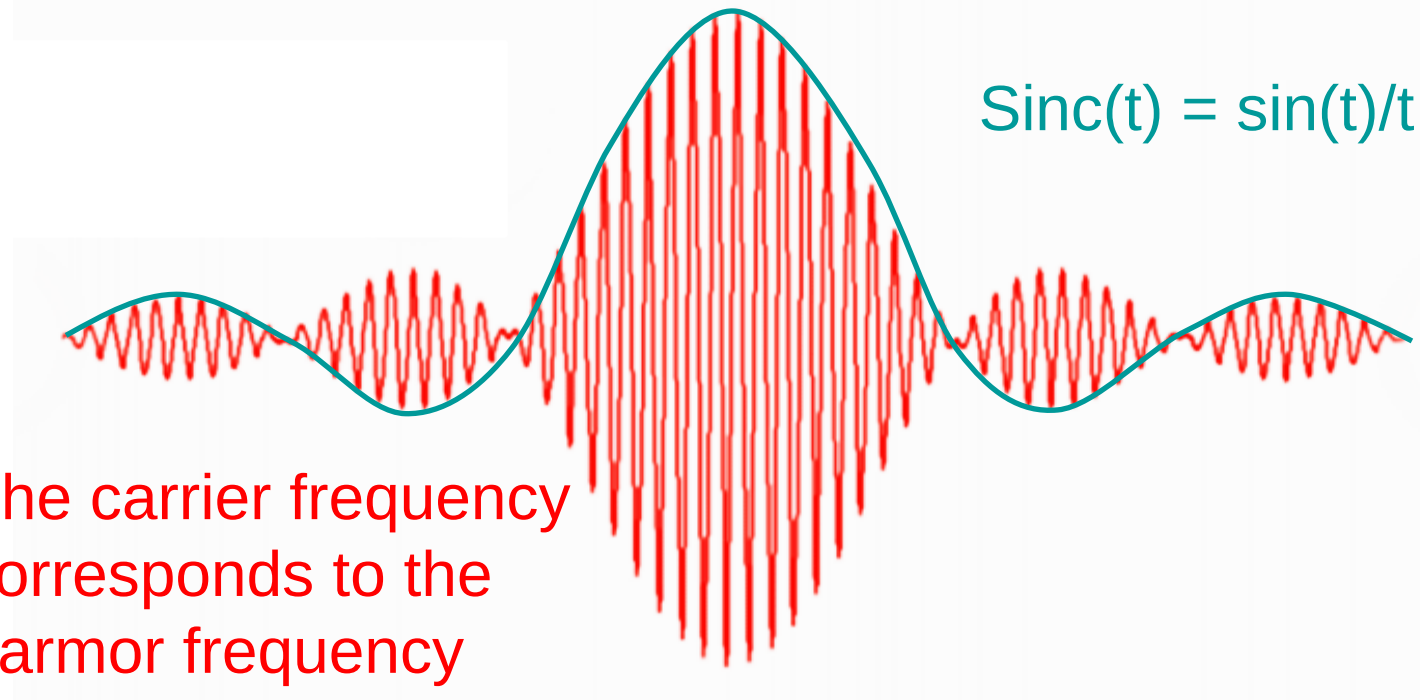
Pitfalls:

- magnetization outside the desired frequency band also contributes to the signal!
- High specific absorption rate for 180°-pulses

Principle of Imaging

Slice selective excitation with a SINC-pulse

Enveloping curve is a (truncated) Sinc function...

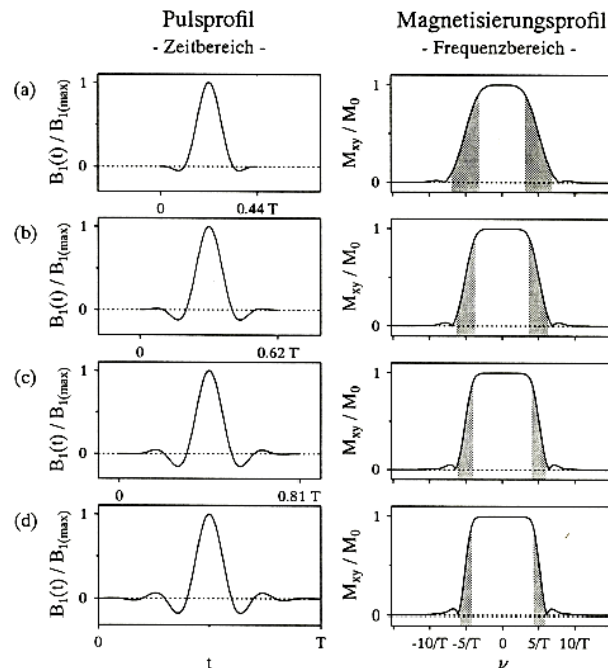


Typical pulse duration is about 1-5 ms

Frequency profiles of different pulses

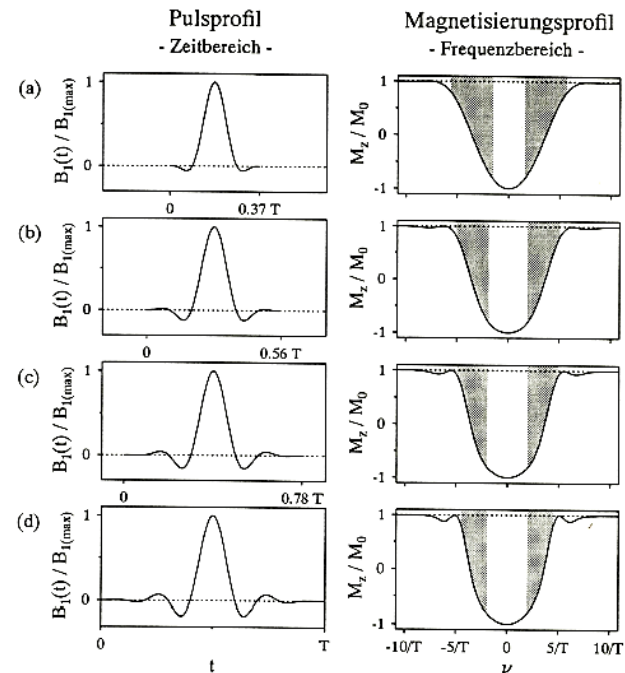
5 FREQUENZSELEKTIVE PULSE

72



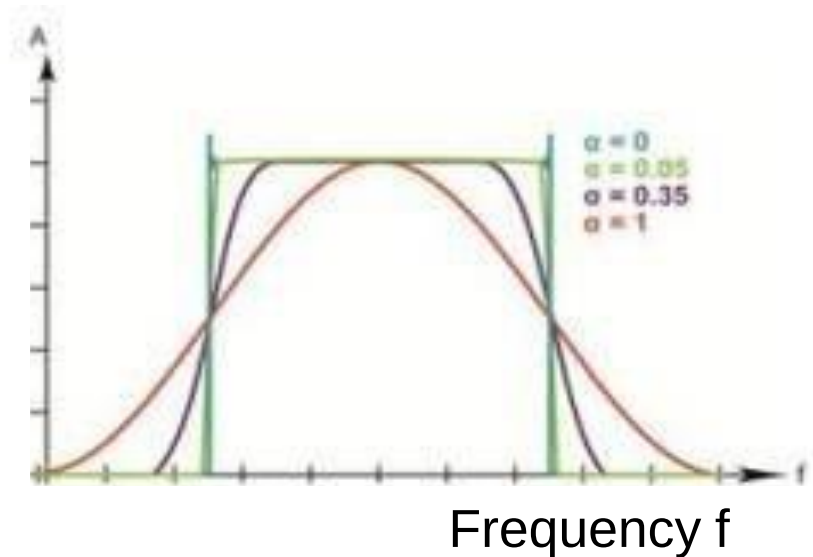
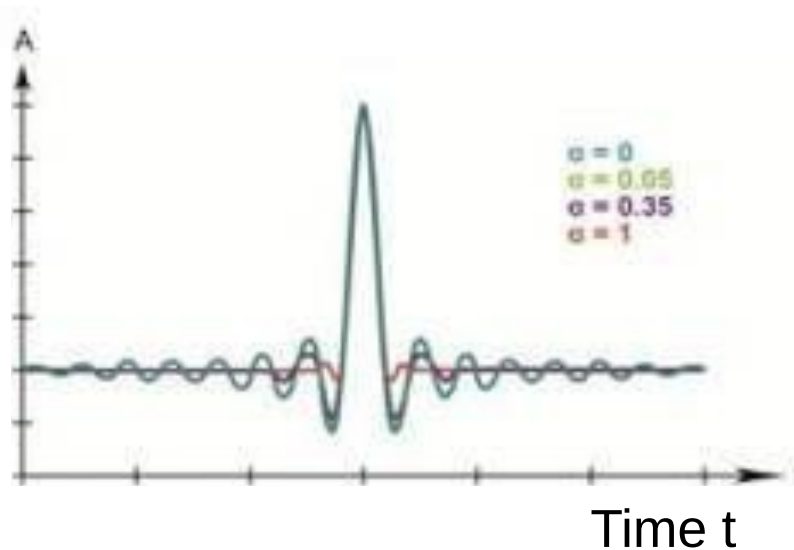
5 FREQUENZSELEKTIVE PULSE

73



Principle of Imaging

Which slice profile is produced by a SINC-pulse?

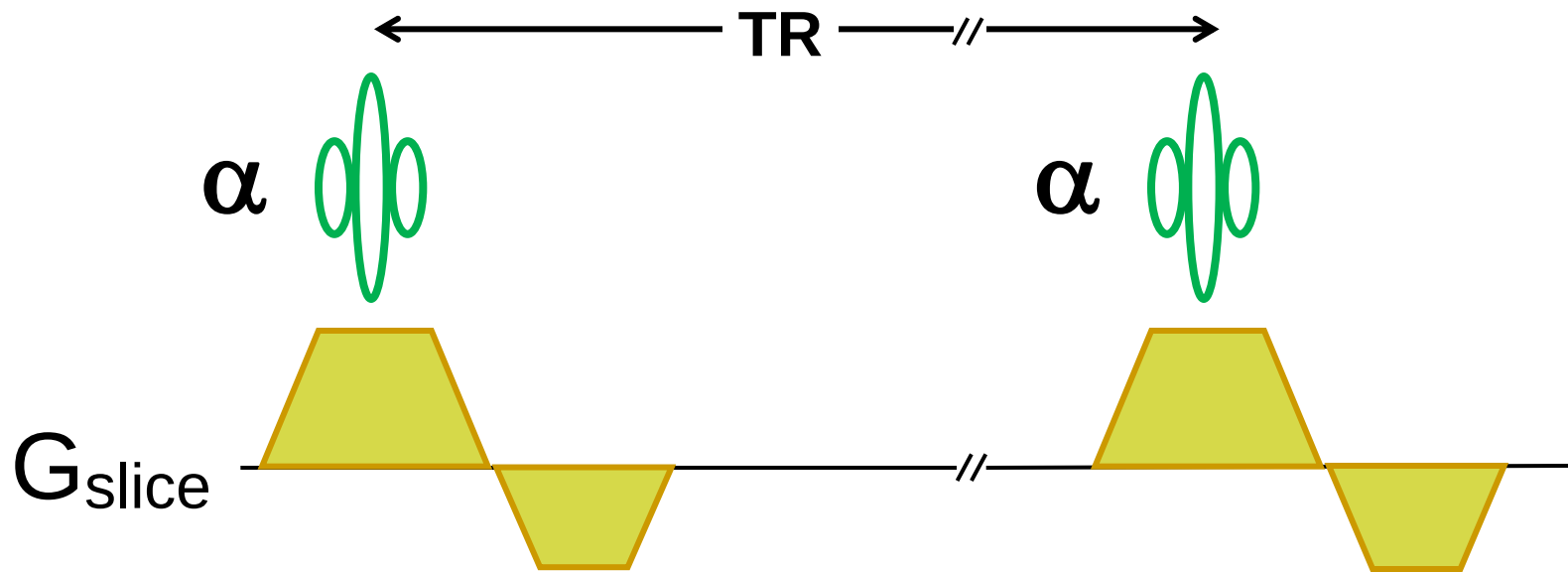


- The more periods the SINC-function considers, the more rectangular is the slice profile.
- For short RF-pulses (strongly truncated, sometimes even without sidebands) rounded slice profiles are produced.

Principle of Imaging

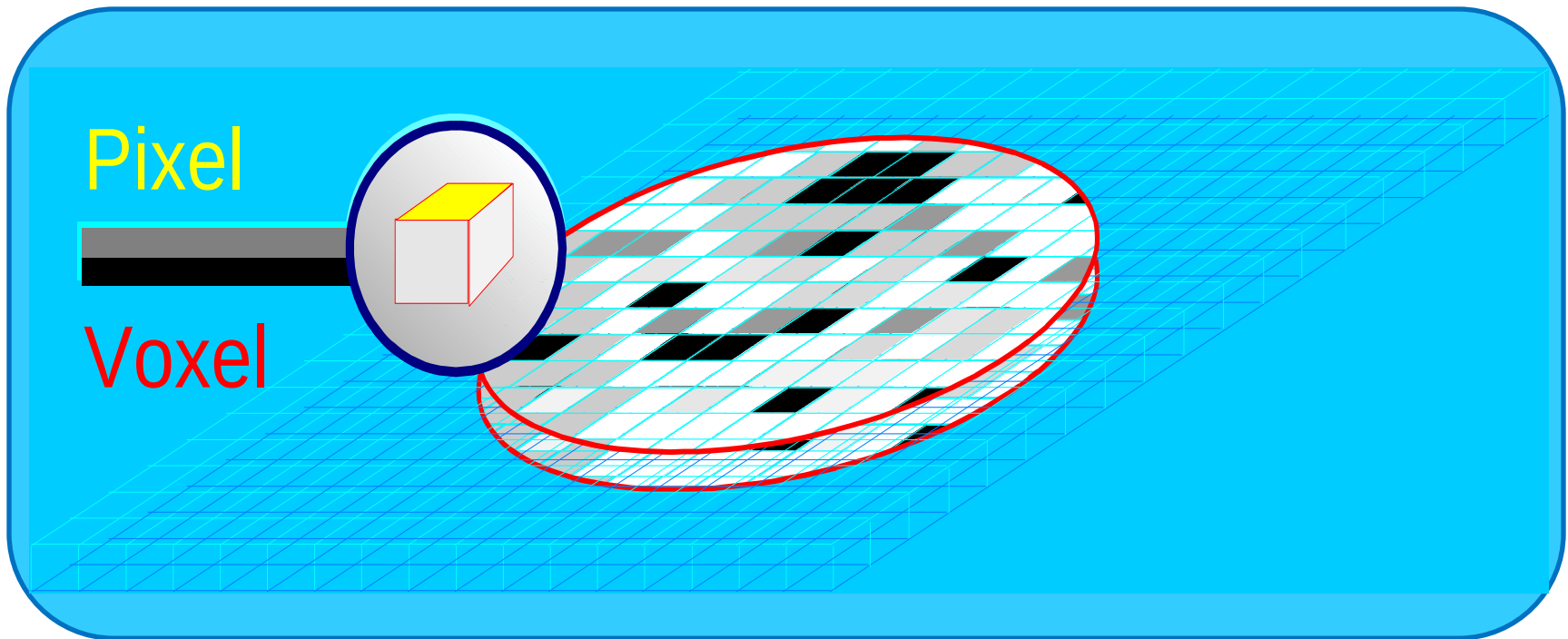
Slice selective excitation

Representation in sequence plot:



Principle of Imaging

Spatial encoding in-plane



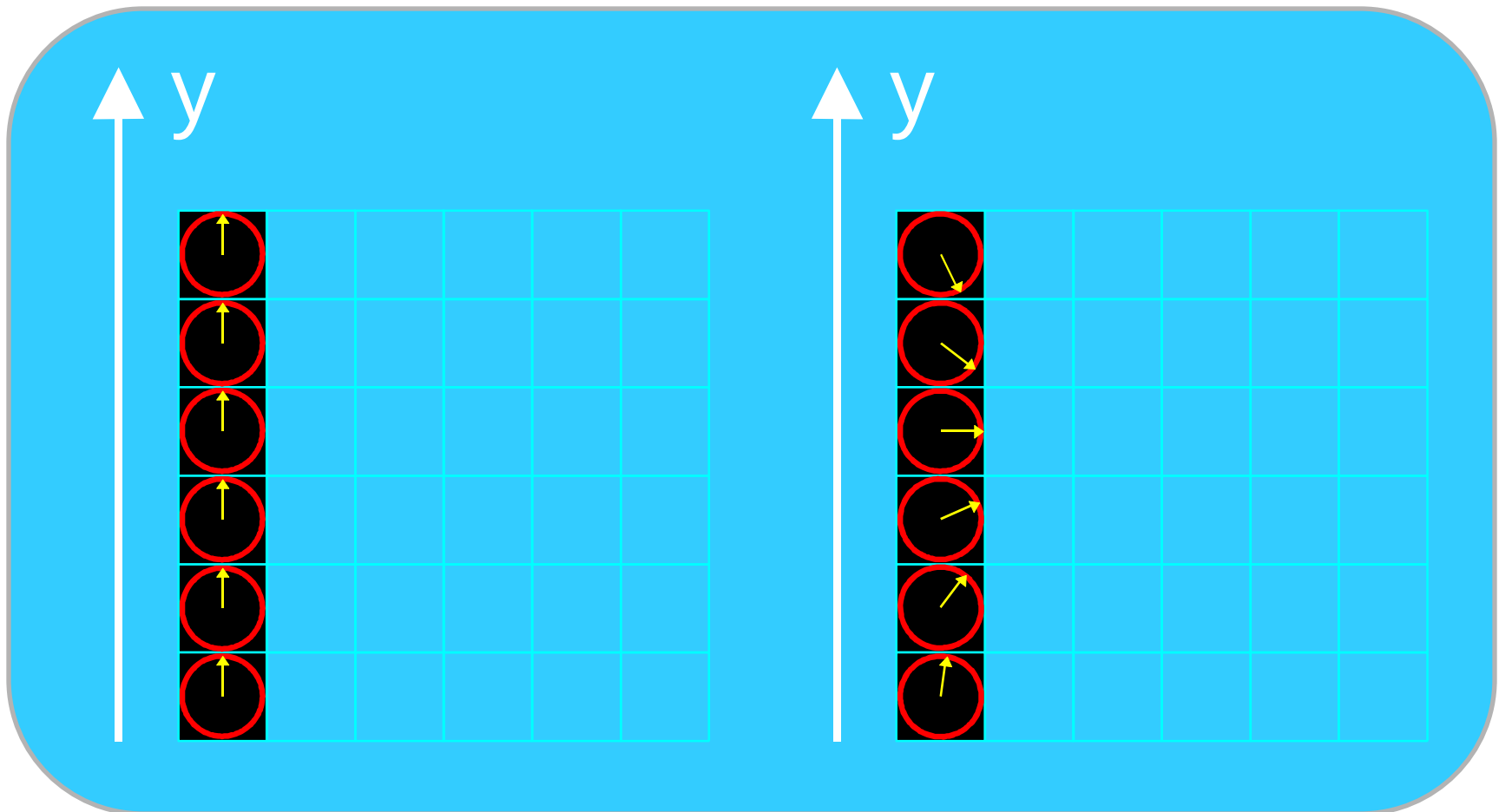
Matrix size: e.g. 256 x 256

Principle of Imaging

Influence of a gradient on signal phase:

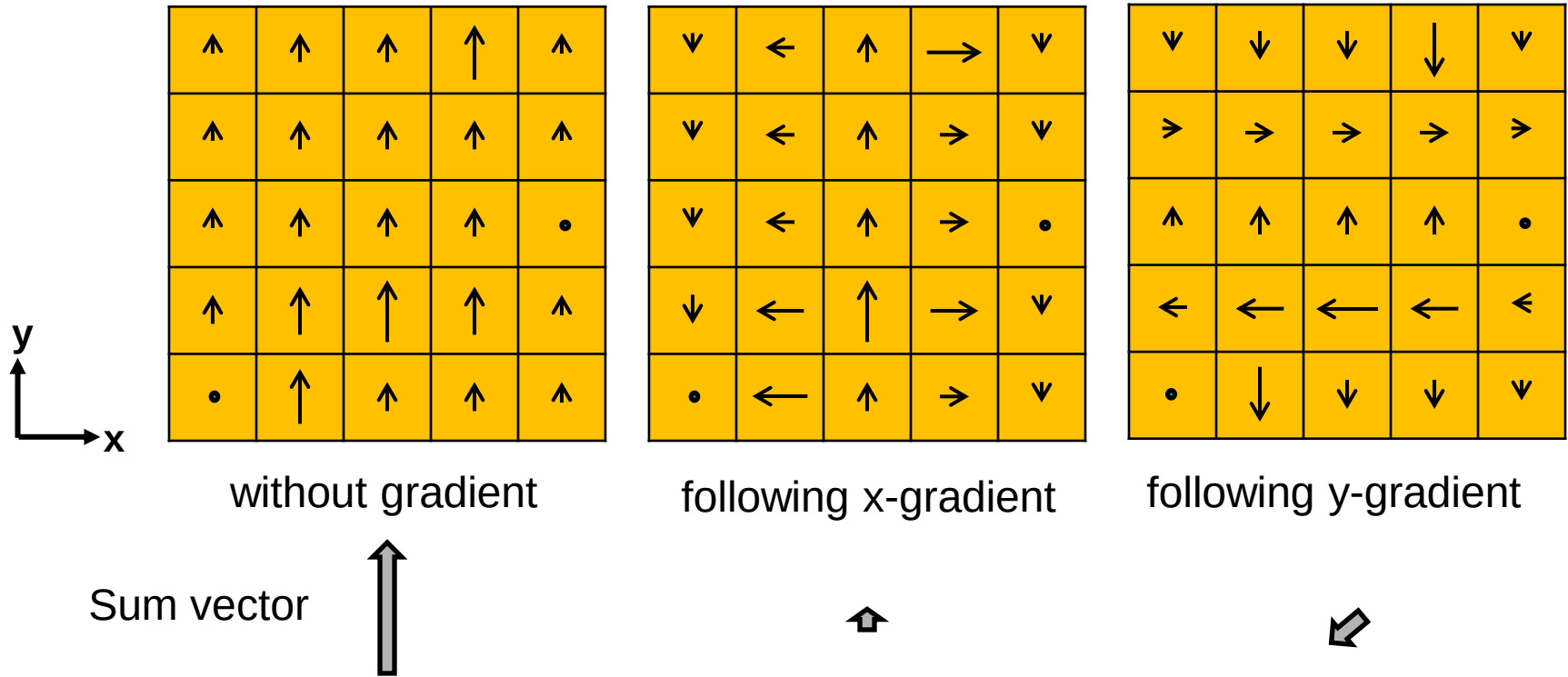
before y-gradient

after y-gradient



Principle of Cartesian Encoding

Encoding of signal contributions from picture elements:

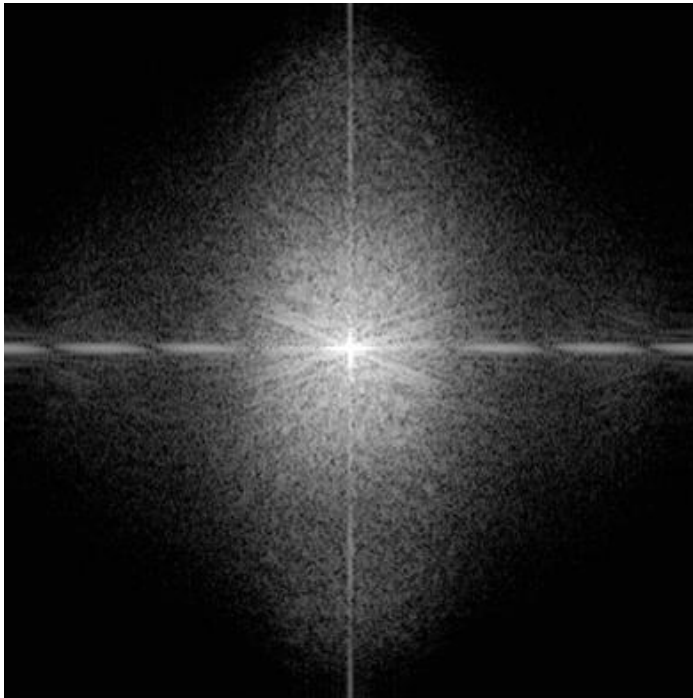


- For unambiguous determination of signal contributions from 5*5 picture elements, recording of the sum vectors of 25 suitable signal superpositions is required.
- Fourier transform is used for analysis.

Principle of Cartesian Encoding

Encoding of signal contributions from the picture elements

Matrix: 256 x 256



Raw data (k-space)

Information about sum
vectors at 65536 different
superpositions

2D-FT

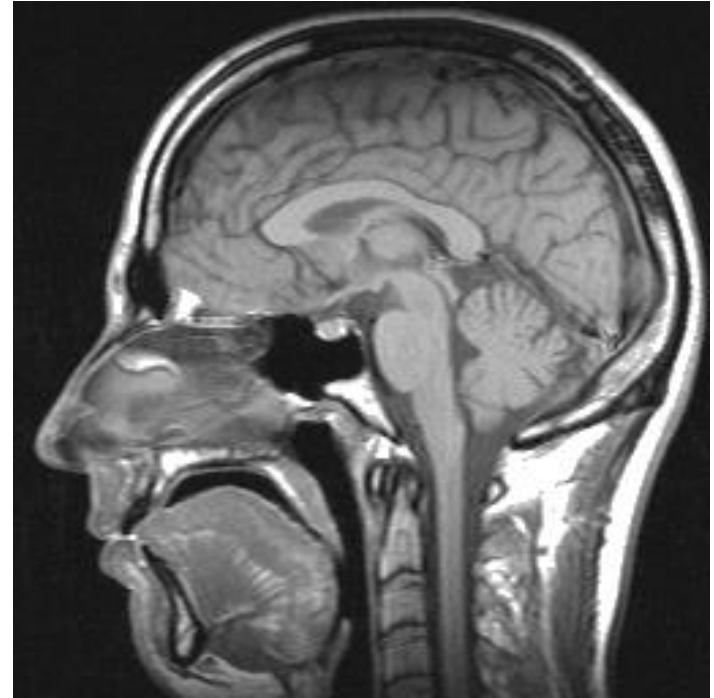


Image data (image space)

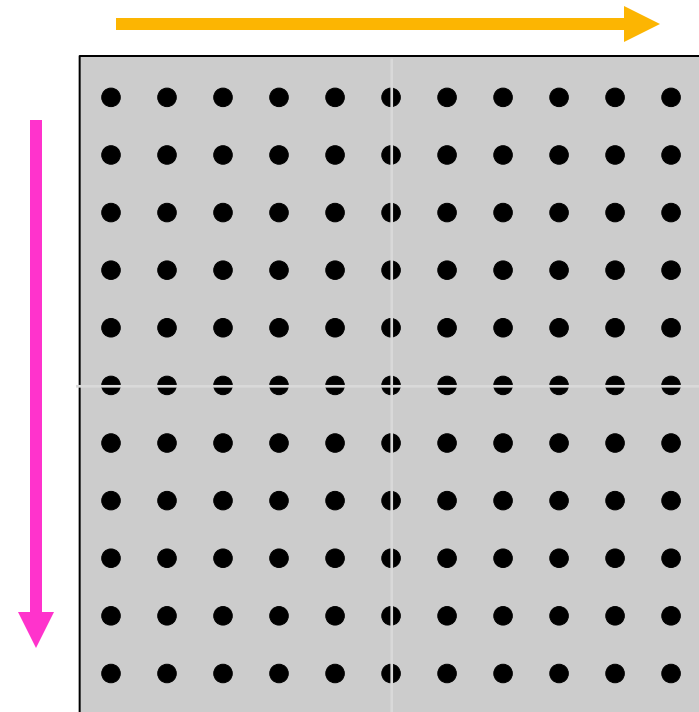
Signal contributions of all 65536
picture elements visualized
as grey values

Principle of Cartesian Encoding

Raw data acquisition:

1. Frequency encoding (read)
2. Phase encoding (phase)

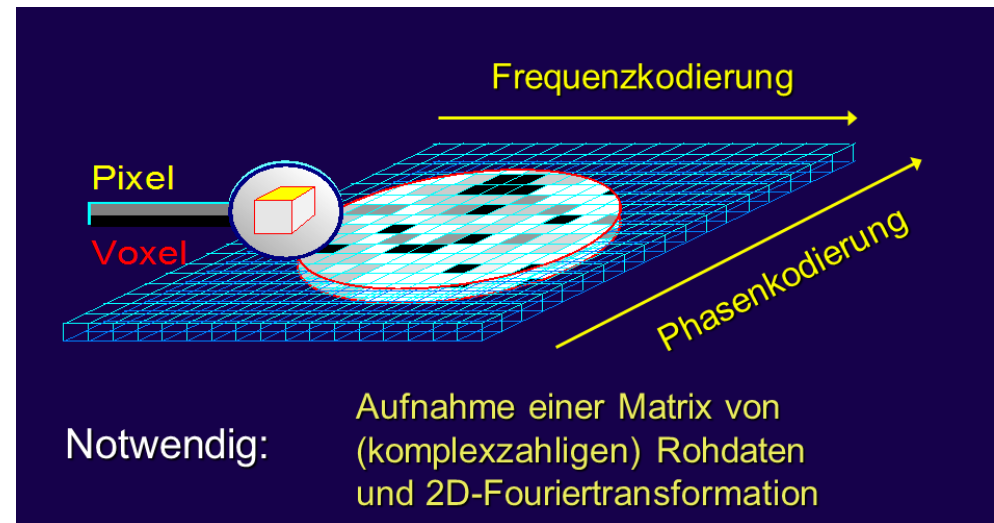
Position of data in k-space



One data point in k-space corresponds to the induced voltage in the RF receiver coil at a time. This voltage corresponds to the sum vector of the transverse magnetization (magnitude and phase angle) at this time.

Principle of Cartesian Encoding

spatial encoding (in-plane)

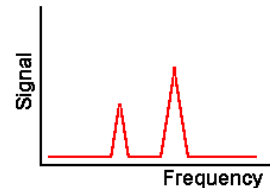
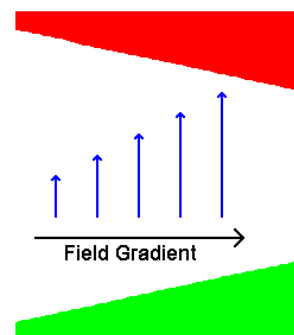
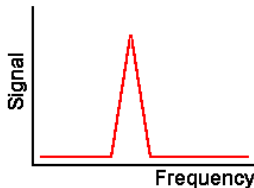
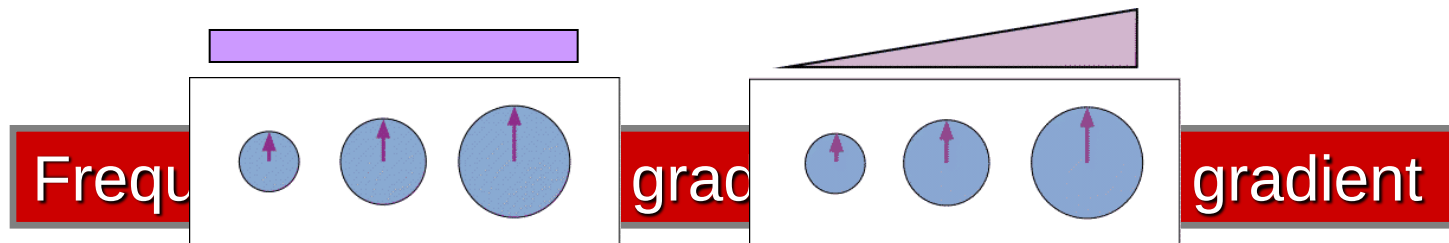


Principle of Cartesian Encoding

frequency encoding

Possibility to assign the signals to a defined space by applying magnetic field gradients

essential: gradient is switched during read-out



Phase encoding

Additional gradient for phase encoding

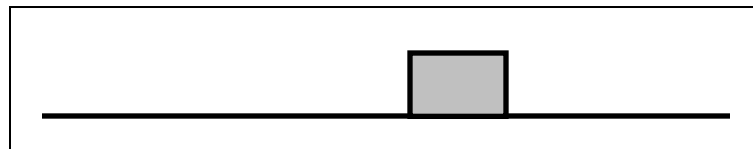
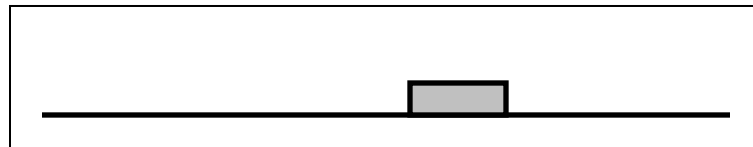
essential: gradient is not switched during excitation or read-out of the signal

spins get a spatial dependent phase

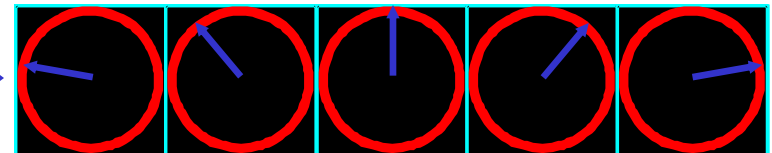
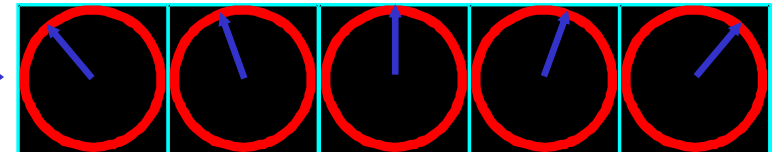
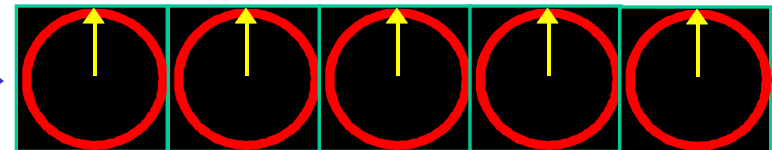
→ Magnetization vectors are oriented in different directions at different places

Principle of Cartesian Encoding

Phase encoding



Phase of magnetization

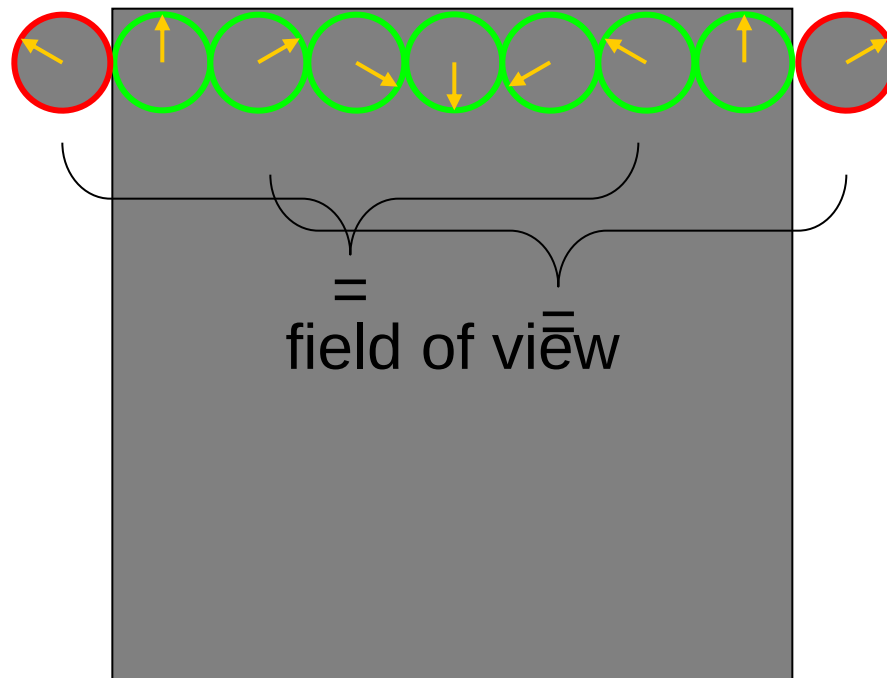


space

A horizontal arrow pointing to the right, indicating the spatial dimension.

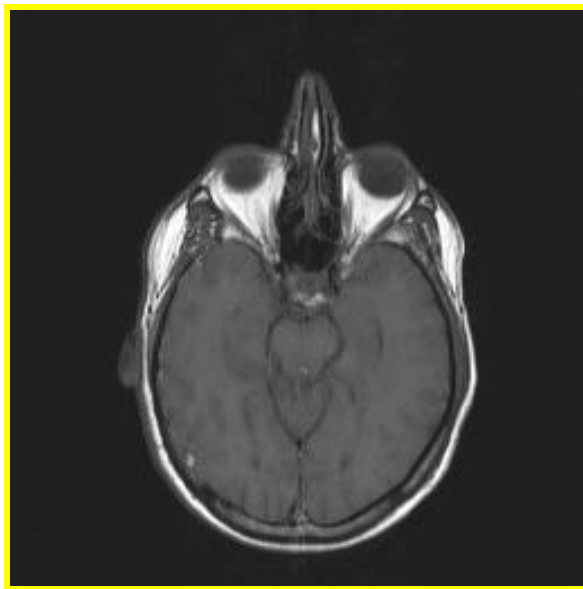
Aliasing

Arise in structures which are in the plane but outside the recorded field-of-view (fold-over) – in phase-encoding direction

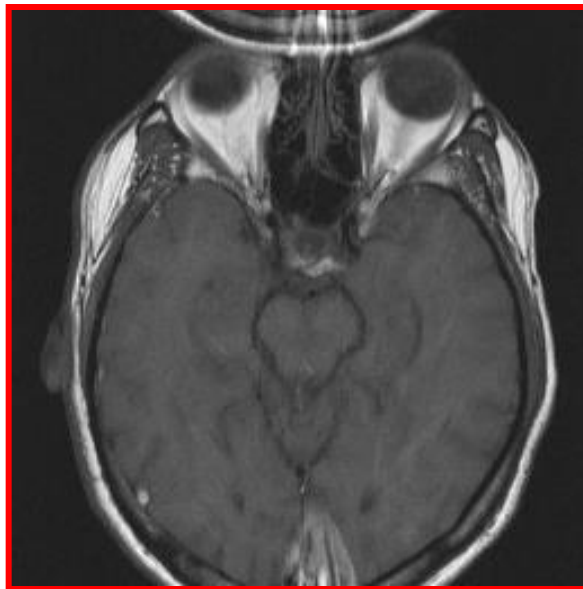


Aliasing

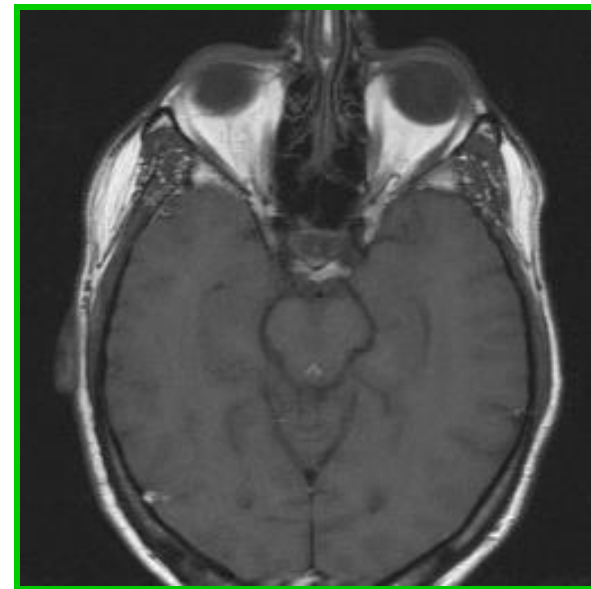
Arise in structures which are in the plane but outside the recorded field-of-view (fold-over) – in phase-encoding direction



Phase enc.: a-p



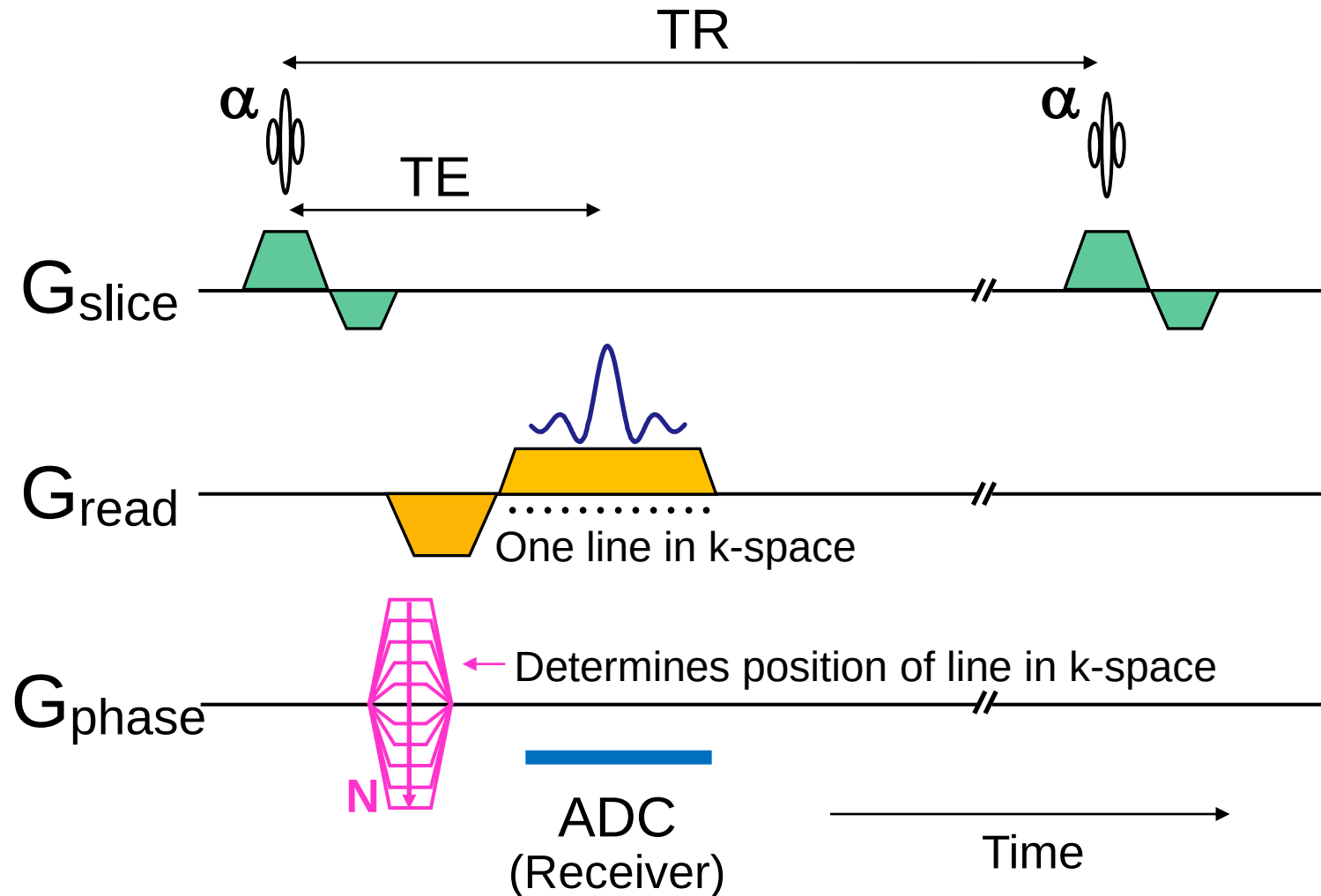
a-p



r-l

Principle of Imaging

Procedure of a gradient echo sequence



Distribution of Information in k-Space

edges, contours, fine structures



rough structures, contrast

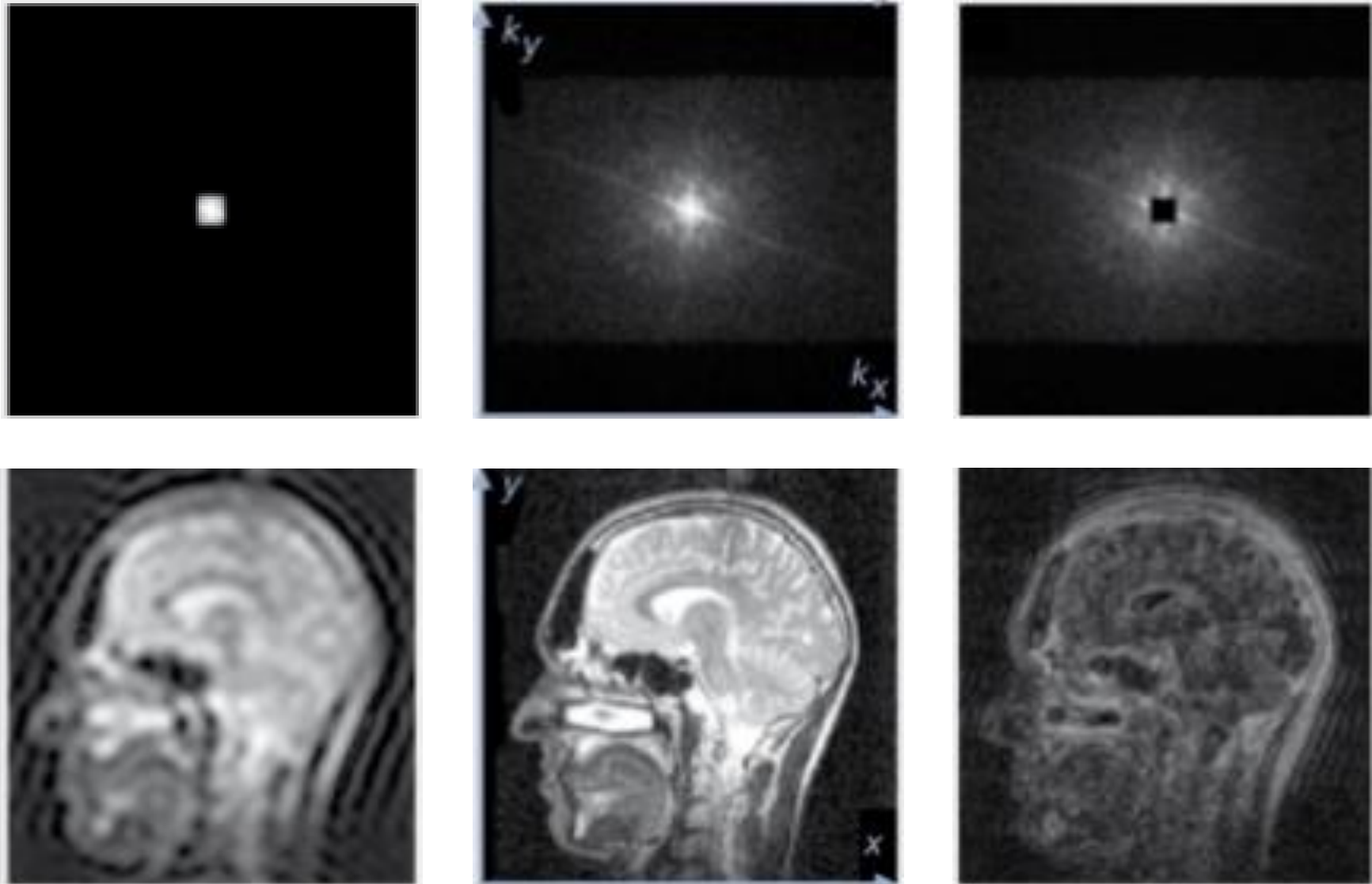
FT
→



Raw data matrix

MR-image

Distribution of Information in k-Space

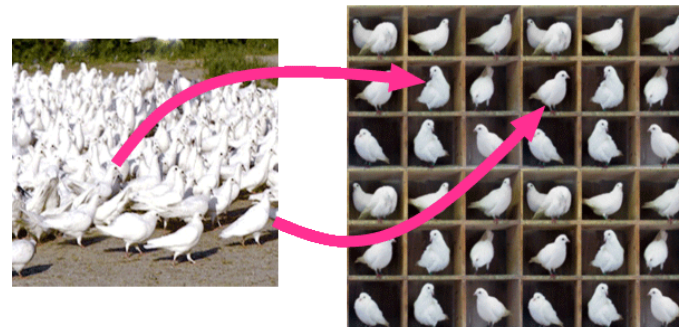


Contrasts are coded in the centre,
edges in the periphery of k-space

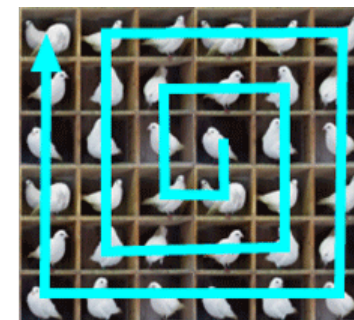
Distribution of Information in k-Space



k-space is an empty room, which has to be filled by digitalized MR-signals in order to calculate the image by Fourier transformation



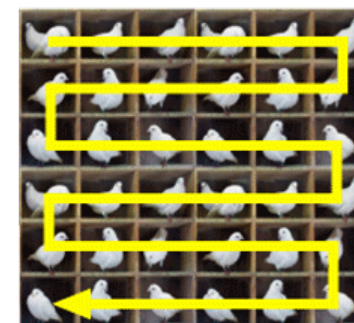
Cartesian



Spiral



Radial

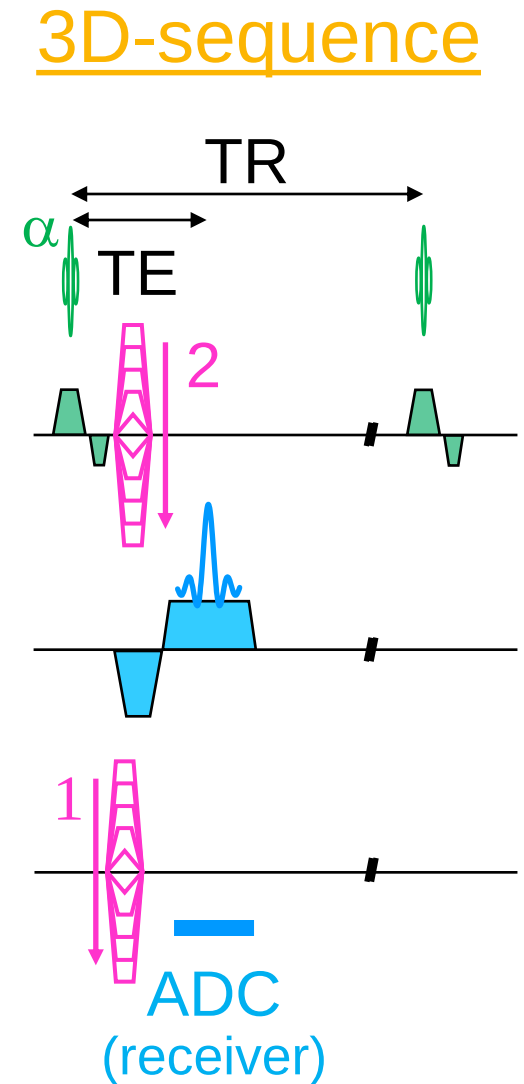
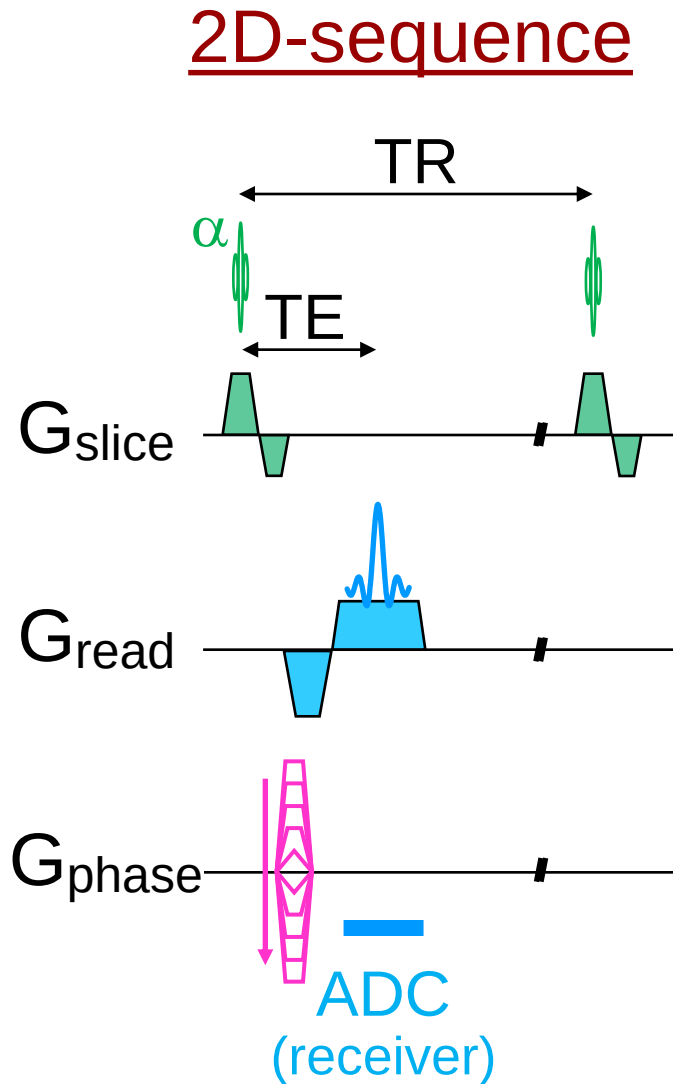
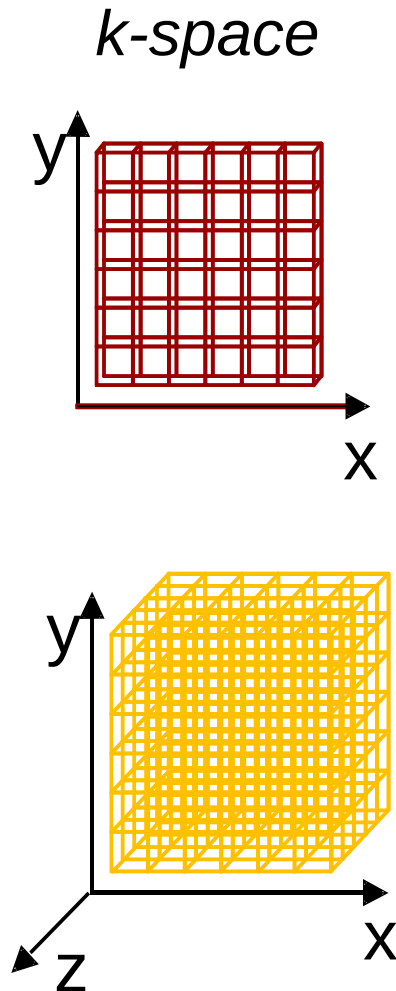


Zig-Zag

<http://mriquestions.com/k-space-trajectories.html>

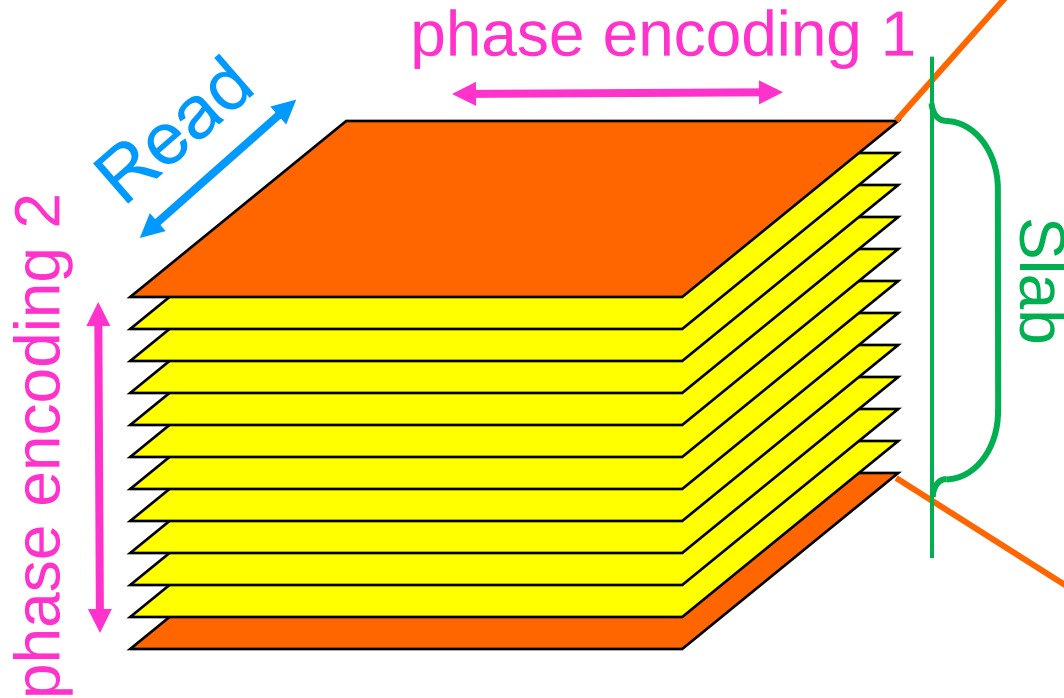
Cartesian Encoding in 2D and 3D Sequences

2D and 3D gradient echo sequences



3D Imaging

- Each RF excitation acts on a thick slab
- Images of single slices are reconstructed by phase encoding in slice direction (phase encoding 2)



Outline

- **Basic principles of magnetic resonance:**
spins, magnetization, RF excitation, relaxation
- **Standard contrasts:**
proton density, T1 and T2 weighting
- **Hardware components of an MR imaging system:**
static magnetic field, gradient fields, RF field T/R
- **Cartesian spatial encoding:**
2D and 3D encoding
- **Standard sequences:**
gradient echo, spin echo, and fast spin echo

Categories of MR Imaging Sequences

Little Sequence Zoo

Standard technique	Gradient echo sequence (GRE)	Spin echo sequence (SE)
Derived faster technique	Echo planar sequence (EPI) used for functional MRI	Fast spin echo sequence (FSE or TSE)

Annotation: Standard technologies acquire only one raw data line in k-space per sequence execution whereas the faster techniques acquire several raw data lines per sequence execution.

Categories of MR Imaging Sequences

What is meant by a “sequence“?

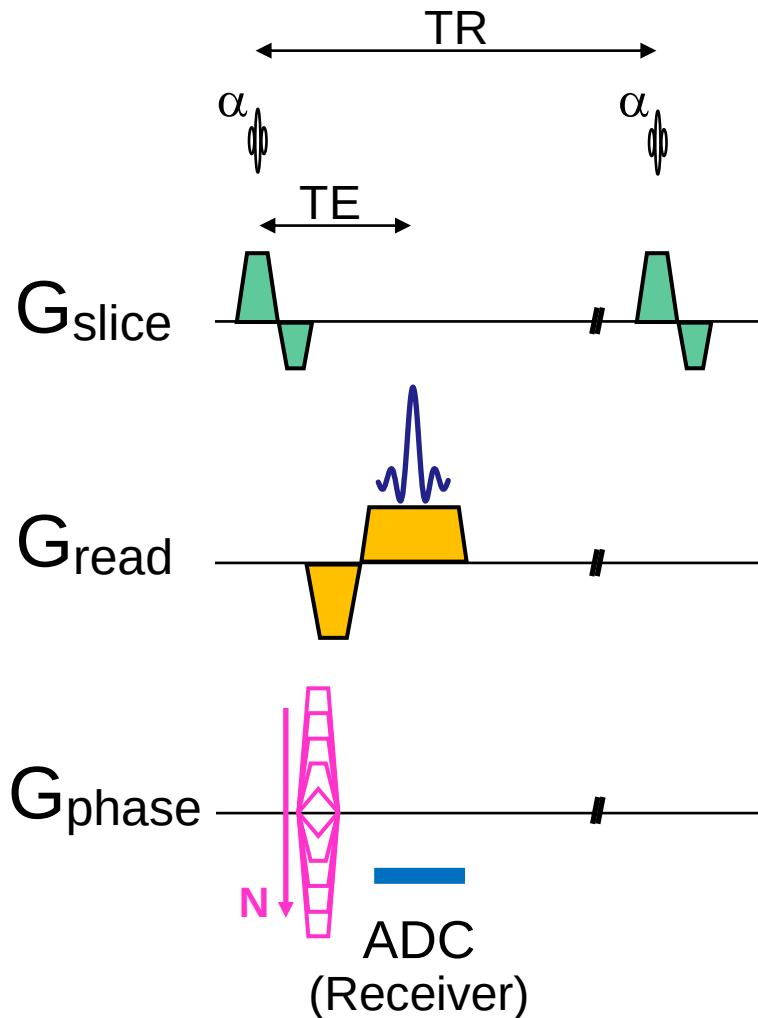
A sequence is a schedule for applications of RF-pulses, gradients and signal readout periods.

If one speaks about a “sequence“ it often refers to the “type of sequence”. For every type of sequence there are characteristic elements which are invariable in the schedule.

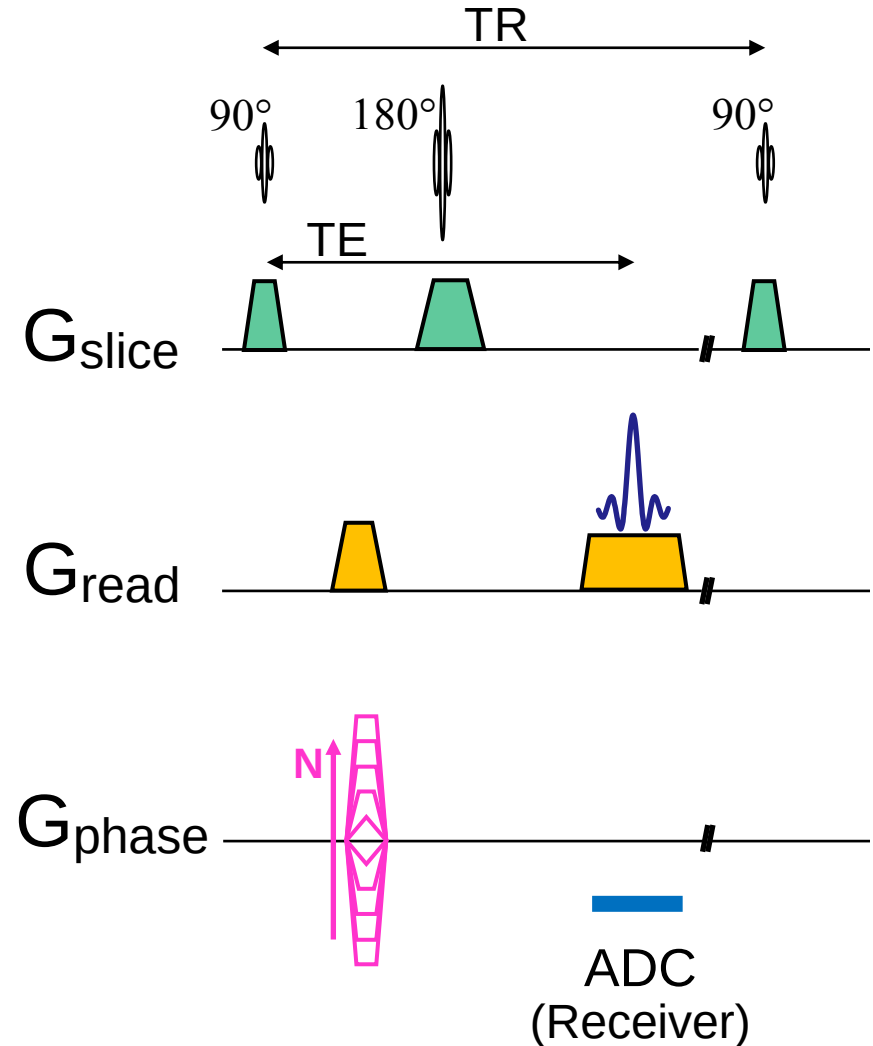
After the selection of the “type of sequence“ many measuring parameters (e.g. TR, TE, slice thickness, matrix size, image field size, ...) can still be changed and adapted. This procedure leads to a modification of the schedule.

Gradient Echo and Spin Echo Sequence

GRE sequence

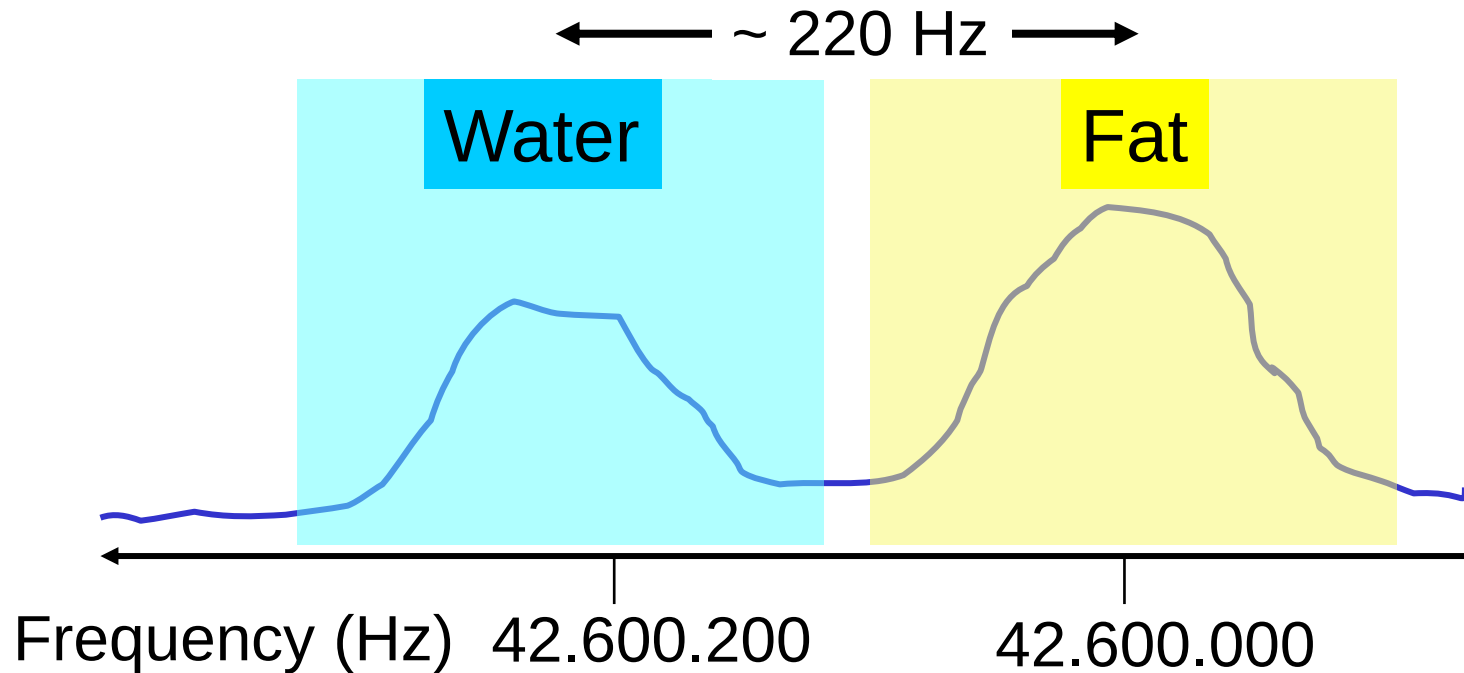


SE sequence



Larmor Frequency Distribution in an Extended Body Area

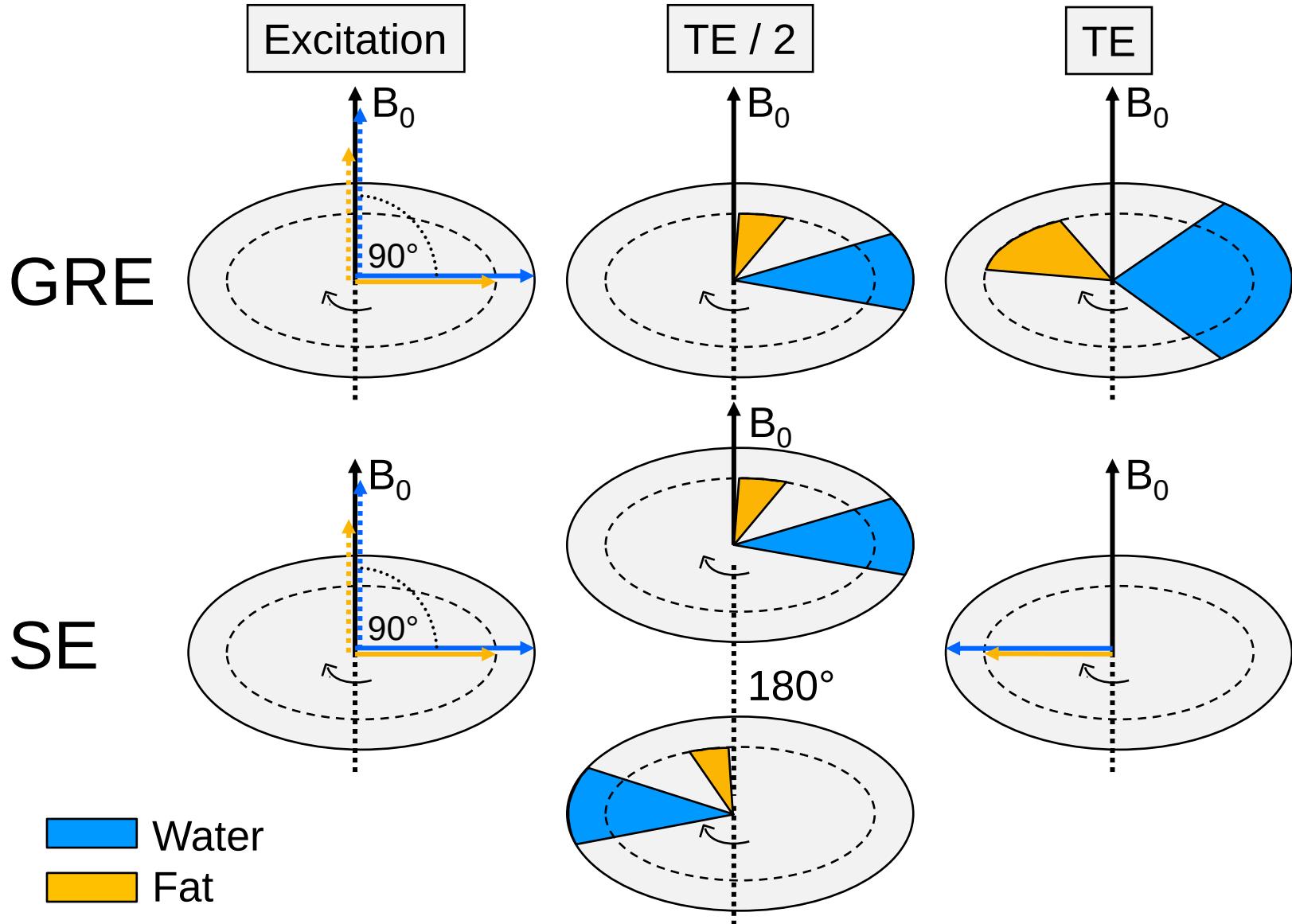
at 1.5 Tesla:



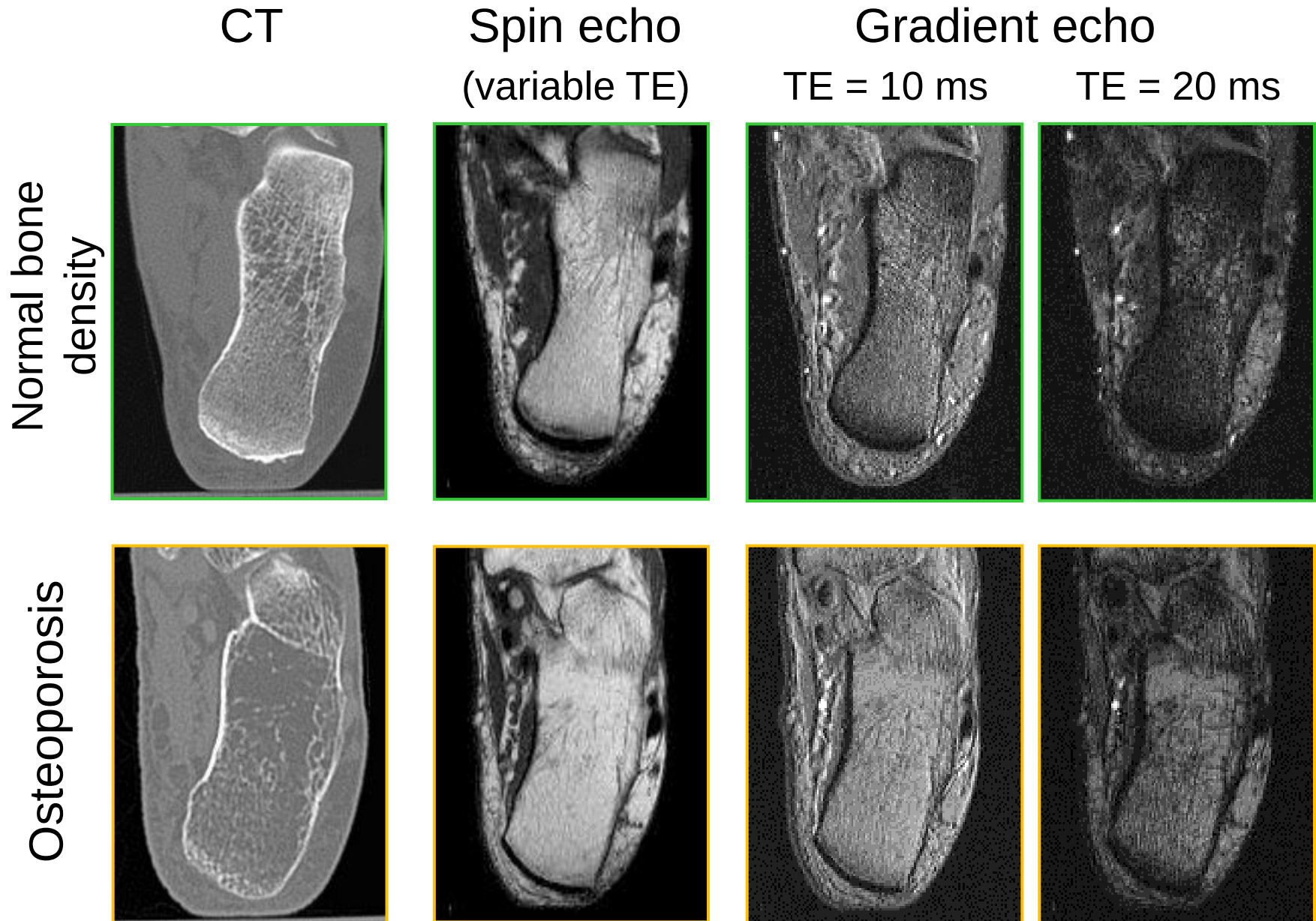
Remember: The difference in frequency between the water signals and the dominant signals of fat amounts to about 3,5 ppm or 150 Hz per Tesla (that is 220 Hz at 1.5 T; 440 Hz at 3.0T)

Gradient Echo and Spin Echo Sequence

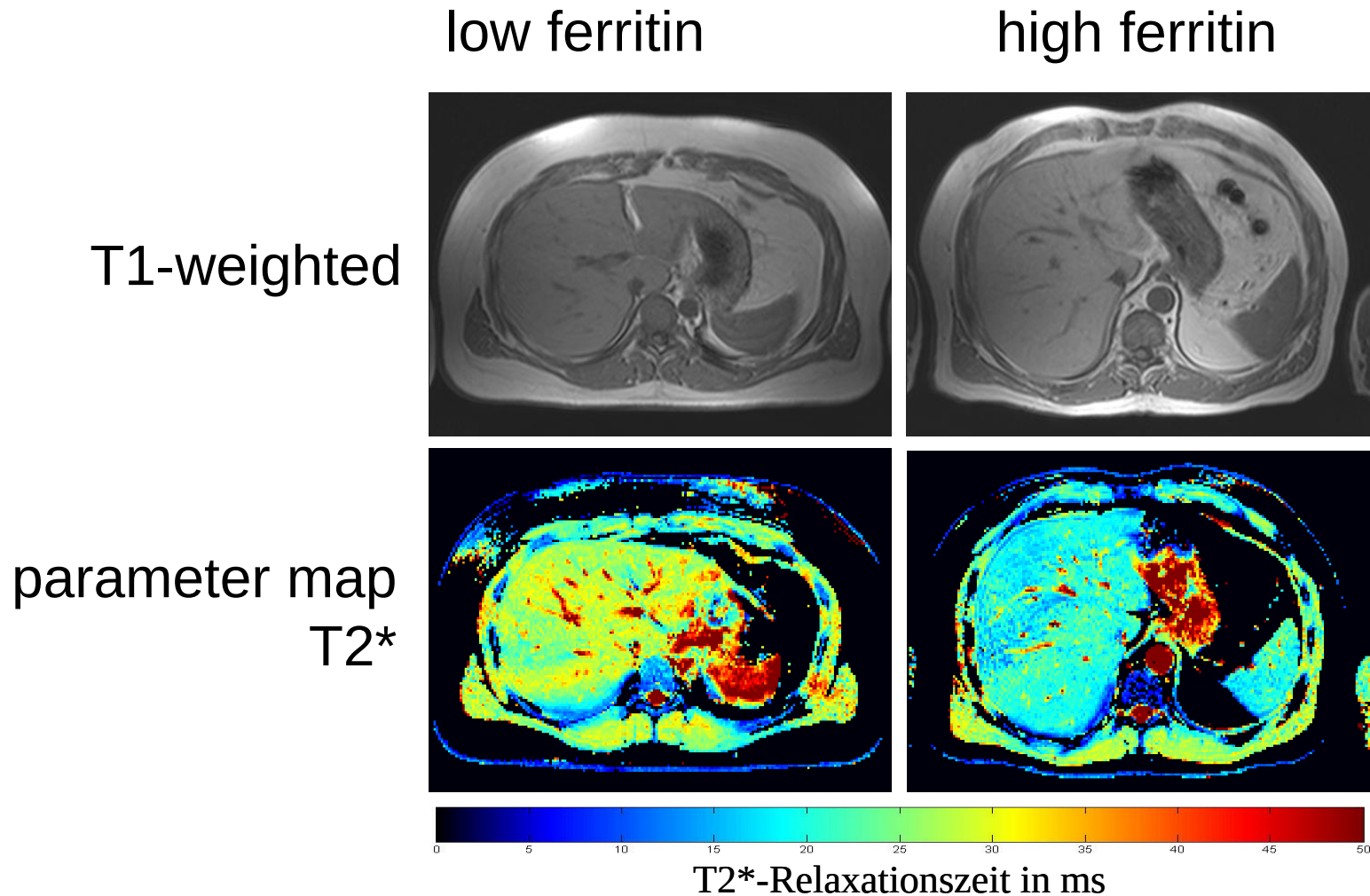
Dephasing in the voxels for GRE and SE sequences



Gradient Echo and Spin Echo Sequence



Gradient Echo and Spin Echo Sequence



Gradient Echo and Spin Echo Sequence

Development of the signal phase

GRE

The phase of the signal components develops according to the Larmor frequencies until readout of signals. The **measured signal strength** corresponds to the **vector sum of the signal components of each pixel**. Magnetic field inhomogeneities, the composition of chemical shift components, and the echo time TE selected significantly influence the measured signal strength. In contrast “real” transverse relaxation (expressed in terms of T2) only plays a minor role.

SE

The **180° pulse** is started halfway through the echo time. It causes an axial reflection of the transverse magnetization and an inversion of the longitudinal magnetization (ideally, the effect at the edge of the slice is in fact too small and the longitudinal magnetization is largely destroyed after the 180° pulse).

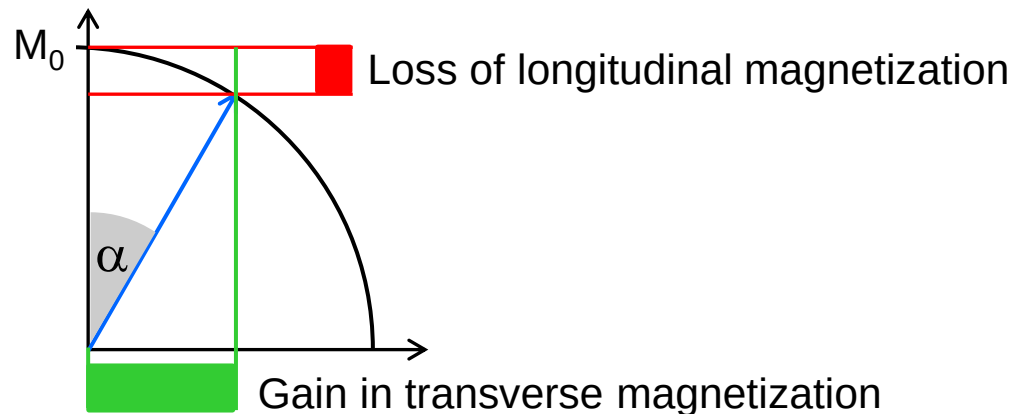
Signal components with different Larmor frequencies are “rephased” and the dependence of the measured signal strength on TE is mainly determined by T2.

Gradient Echo and Spin Echo Sequence

Selection of flip angle

GRE

If smaller flip angles than 90° are used, longitudinal magnetization remains available immediately after the RF-excitation. Therefore, in this case it is possible to use shorter TR than for SE sequences. The highest signal yield per measurement period is achieved by excitation with the Ernst angle $\alpha_{\text{Ernst}} = \arccos \exp(-TR/T1)$. Example: For a tissue with $T1 = 1\text{ s}$ and the application of a (spoiled) GRE sequence with $TR = 100\text{ ms}$ a flip angle of 25° leads to the highest signal yield.



SE

The 180° refocusing pulse leads to a largely destroyed longitudinal magnetization. Therefore, the highest signal yield is always achieved by applying the pulse sequence: $90^\circ - 180^\circ$.

Gradient Echo and Spin Echo Sequence

GRE

- Excitation with a slice selective RF pulse with 5° - 90° flip angle
- Short repetition times possible, with relatively high signal yield at small flip angles
- No rephasing of effects of field inhomogeneities and different chemical shift components within the voxels
→ T_2^* weighting und opposed-phase imaging are possible
- Short minimal echo times can be realized

SE

- Excitation by a slice selective pulse sequence: 90° - 180°
- Rephasing of field inhomogeneities and different chemical shifts within the voxels are always guaranteed
→ T_2^* weighting und opposed-phase imaging are impossible
- Prolonged minimal echo times due to the additional 180° pulse
- For reasonable signal yield relatively long repetition times (> 300 ms) are necessary, since longitudinal magnetization is virtually zero after signal recording

Gradient Echo and Spin Echo Sequence

Possible contrasts and measuring parameters

GRE

T1-weighting: is controlled by TR **and** the excitation flip angle
T1-weighting gets stronger for shorter TR and/or higher excitation flip angles

T2*-weighting: is controlled by TE
T2*-weighting gets stronger for longer TE (eg, 15-20 ms)
(Cave! Phase relation between water and fat signals matters as well)

T2-weighting: impossible!!

SE

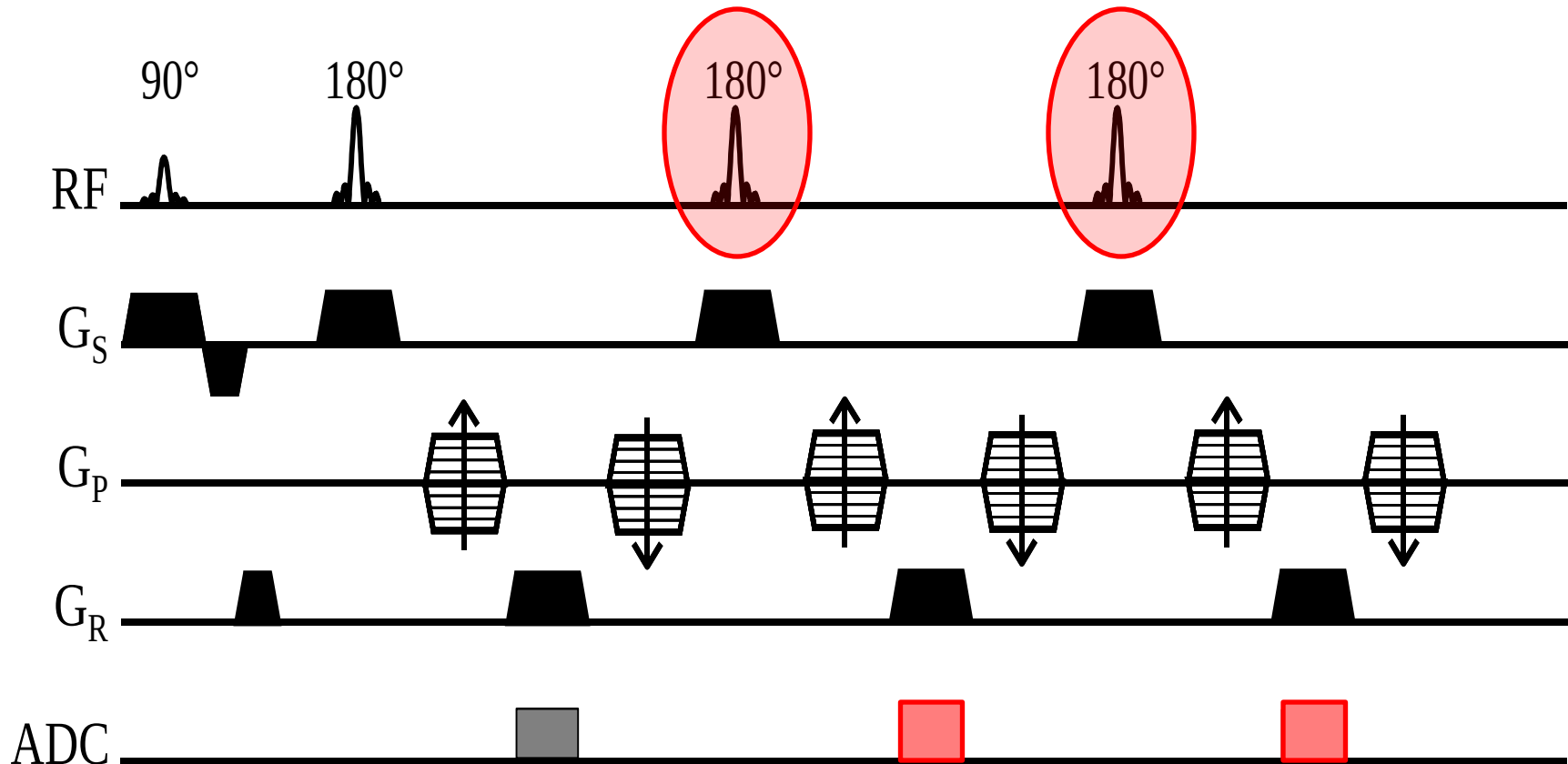
T1-weighting: is controlled by TR
T1-weighting gets stronger for shorter TR (eg, < 800 ms)

T2*-weighting: impossible!!

T2-weighting: is controlled by TE
T2-weighting gets stronger for longer TE (eg, 100 ms)

Acceleration of Spin Echo Recordings

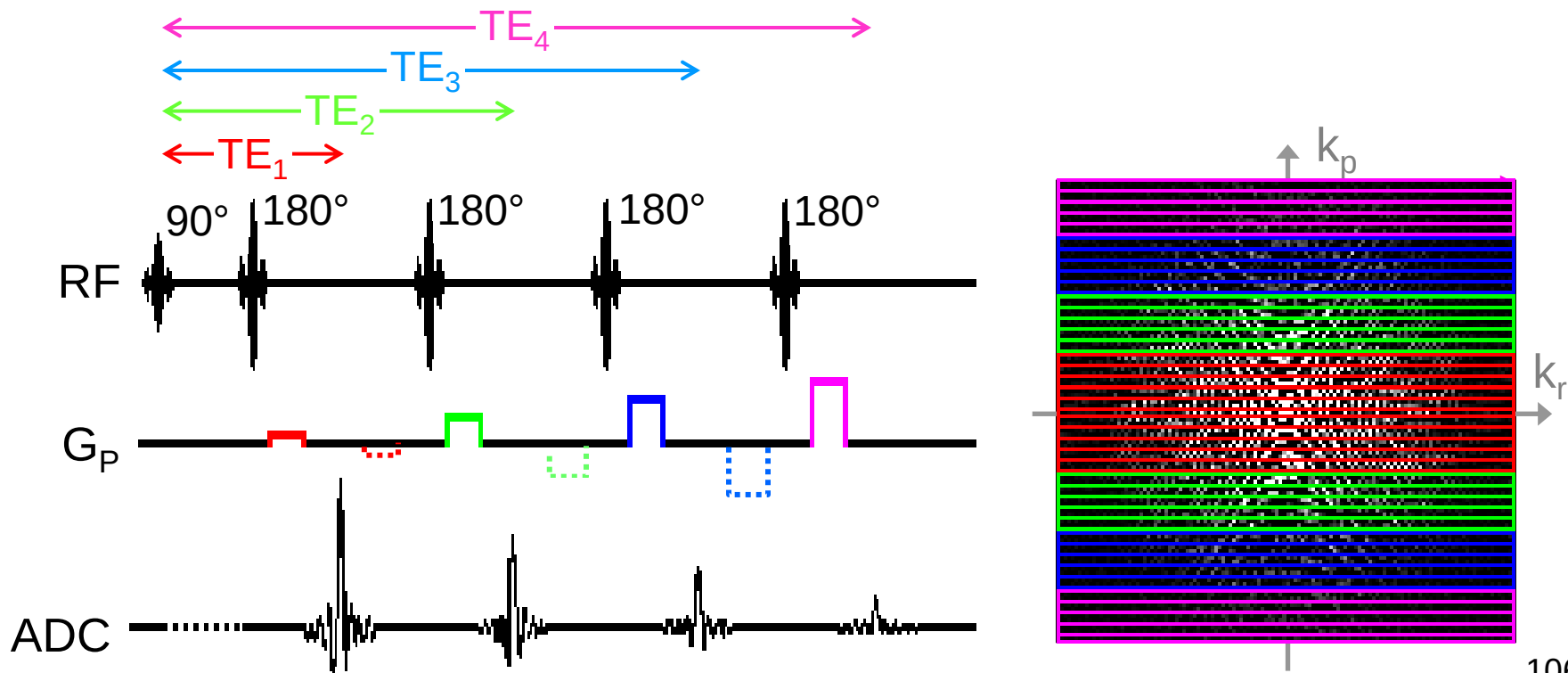
Fast spin echo (FSE) = Turbo spin echo (TSE)



Following a 90° excitation pulse and a 180° refocussing pulse further 180° pulses are applied and further raw data are recorded.

Fast Spin Echo (FSE) Sequences

- Different raw data lines are recorded **at different echo times**.
- Raw data recorded at the same echo time are arranged together in k-space.
- The echo time steps are minimized in k-space in order to reduce related artifacts.
- The **echo time** which is used to fill the **central area in k-space** gets stated for the recording (in this case TE_1).



Fast Spin Echo (FSE) Sequences

- Since several spin echoes (i.e. multiple raw data lines for different phase encoding steps) are recorded after each 90° excitation, the **measuring time is shortened considerably**.
- The **number of spin echoes recorded after a 90° excitation** in a FSE sequence is referred to as **echo train length (ETL)** or **turbo factor**.
- In an FSE sequence a minimum of 3 echoes and a maximum of the entire k-space are recorded after a 90° excitation. If the entire k-space is recorded within one echo train one speaks of a single shot FSE sequence, e.g. HASTE (= Half fourier-Acquired Single shot Turbo spin Echo)
- The single spin echoes of an echo train are always read out using a **fixed interval** in-between, which is referred to as “**echo spacing**”.
- The duration of the echo train for one slice is slightly longer than the product of “ETL · echo spacing” (important for the maximum number of slices in nested multi-slice imaging).

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Fast Spin Echo (FSE) Sequences

Comparison: SE and FSE (ETL = 7) images



SE image

TR = 300 ms, TE = 15 ms

Recording time 2:37 min

Multi-slice imaging with a
maximum of 16 slices



FSE-7 image (ETL:7)

TR = 300 ms, TE = 15 ms

Recording time: 23 sec

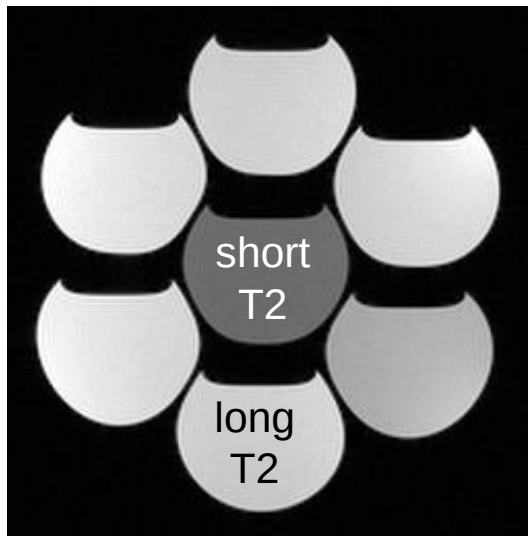
Multi-slice imaging with a
maximum of 2 slices

Fast Spin Echo (FSE) Sequences

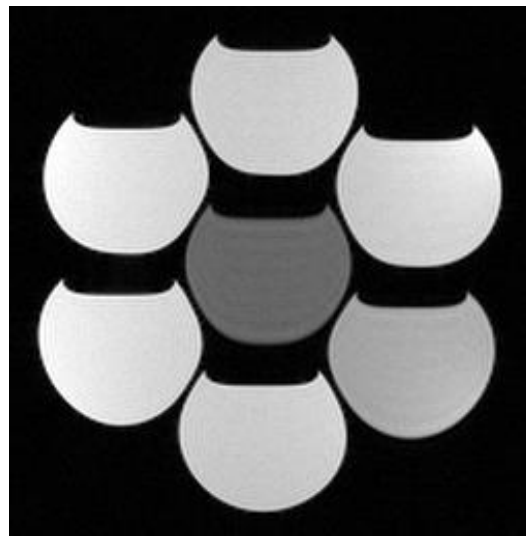
Since raw data lines are recorded at **different echo times**, **artifacts** occur in **phase encoding direction**.

The artifacts are the more pronounced

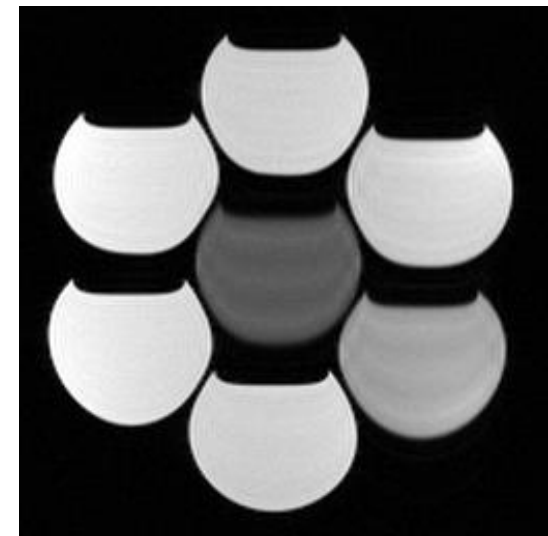
- the **longer the echo train length** is
- the **shorter the relaxation time T2** of the recorded material is



SE
TR = 3 s, TE = 15 ms
TA = 12:52 min



TSE-3
TR = 3 s, TE = 15 ms
TA = 4:19 min



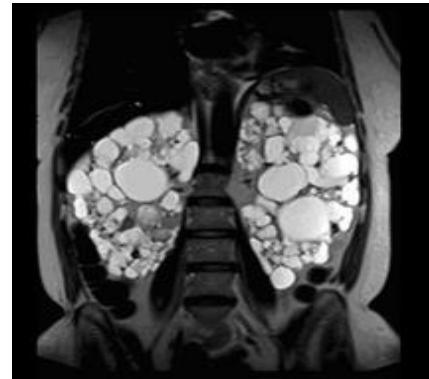
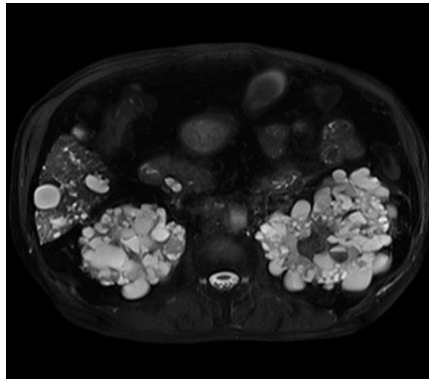
TSE-7
TR = 3 s, TE = 15 ms
TA = 1:52 min

HASTE

(dis)advantages

- + very short acquisition time (< 1 s), insensitive to patient movement or breathing
- tissue with inherent short T2 gives no signal at the end of the pulse train

broad field of applications in clinical routine



Fast Spin Echo (FSE) Sequences

Advantages and disadvantages of FSE sequences in comparison to SE sequences:

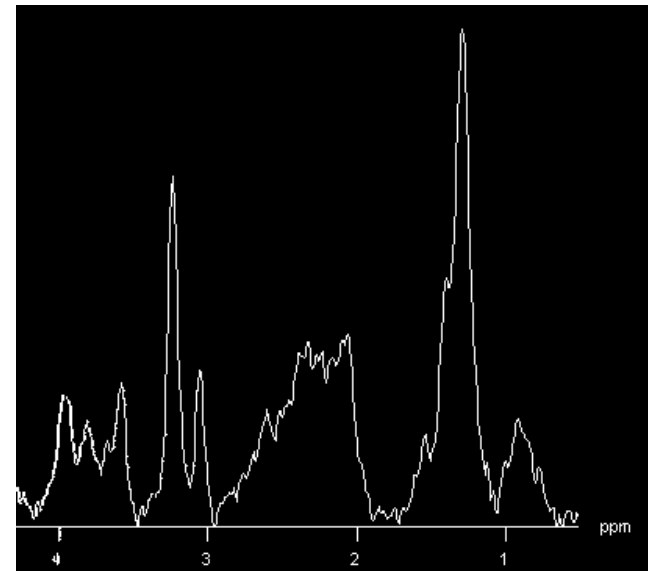
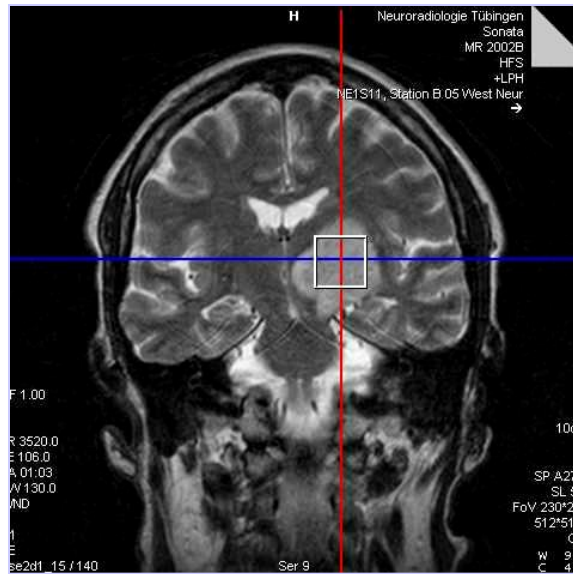
Advantages:

- + Significantly shorter measuring times for T2-weighted images, only a few minutes measurement time for high-resolution T2-weighted images
- + Spin echo contrast (especially T2-contrast) becomes feasible in breath-hold exams. With single shot sequences even restless patients can be well examined (e.g. using HASTE, recording time < 0,5 s per image)

Disadvantages:

- Tissues with short T2 are often blurred
- Poor detectability of smallest lesions (e.g. cerebral cortex, meniscus) caused by FSE artifacts
- Lower maximum number of slices in multi-slice imaging
- Fat signal is slightly enhanced compared to spin-echo recordings (at same TR and TE)
- High energy load due to many 180° pulses (SAR limit often reached)

MRI vs. MRS



Chemical shift

Why we can separate signals from identical nuclei...

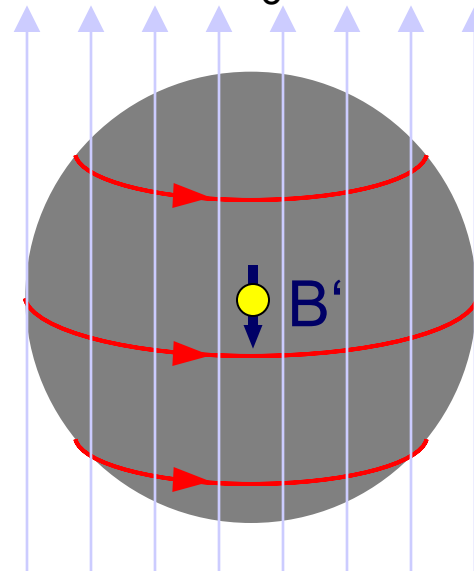
Larmor frequency of a spin system is always:

$$\omega = \gamma \cdot B$$

Gyromagn. ratio γ :
Prop.-factor between spin
and magnetic moment

but: field strength B encountered by the nucleus is not the one given by the applied static magnetic field B_0 !

Chemical surrounding causes a shift of the Larmor frequency due to the shielding of the electron hull



induced e⁻-movement

Induced counterfield

$$B_{\text{eff}} = B_0 - B' = B_0 \cdot (1 - \sigma)$$

σ : shielding constant

Chemical shift

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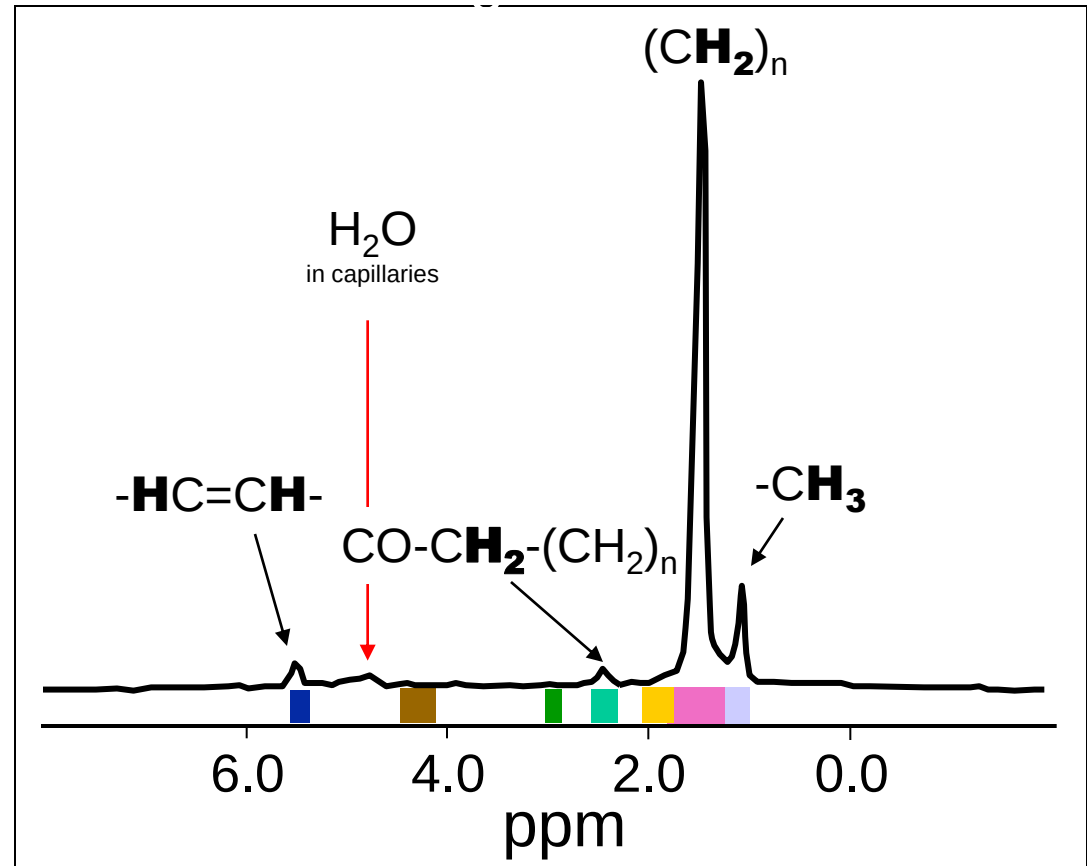
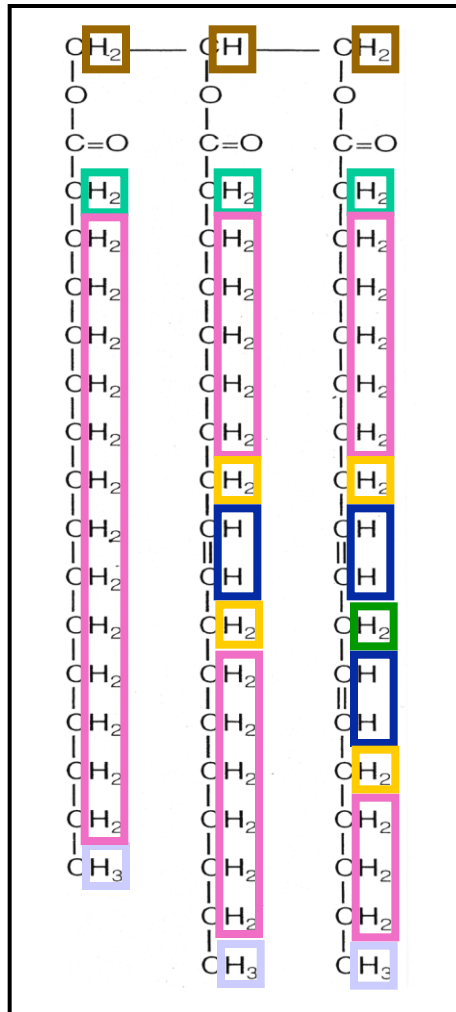
Change in B by shielding is very small – thus, the change of the resonance frequency is very low

- in the scale of 10^{-6} (millionth)
- given by chemical shift (δ)
in **p**arts **p**er **m**illion (ppm)
- standard: Tetramethylsilan $[\text{Si}(\text{CH}_3)_4]$ at 0 ppm
(water at 4.7 ppm, creatine at 3.05 ppm, methylene at 1.3 ppm)
- independent on field strength

$$\delta = \frac{\nu - \nu_{\text{ref}}}{\nu_{\text{ref}}}$$

Chemical shift

Why we can separate signals from identical nuclei...



¹H spectrum from yellow bone marrow
triglyceride

MRS basics

In-vitro:

- One of the most important analytical methods for determination of structure and dynamics of and interplay between molecules

In-vivo:

- Low concentrated metabolites, which cannot be visualized by MRI, are detectable
- selective examination of small volumes of interest (few cm³)
- Non-invasive assesement of metabolism in healthy and diseased tissue

MRS basics

Imaging:

- Signal intensities from many picture elements are recorded simultaneously and are stored in a matrix – k-space (256^2 , 512^2)
- Spatial encoding by gradient during read-out
- Contrast in the image depends on different tissue types and chosen sequence parameters (TE, TR, ...)

Spectroscopy:

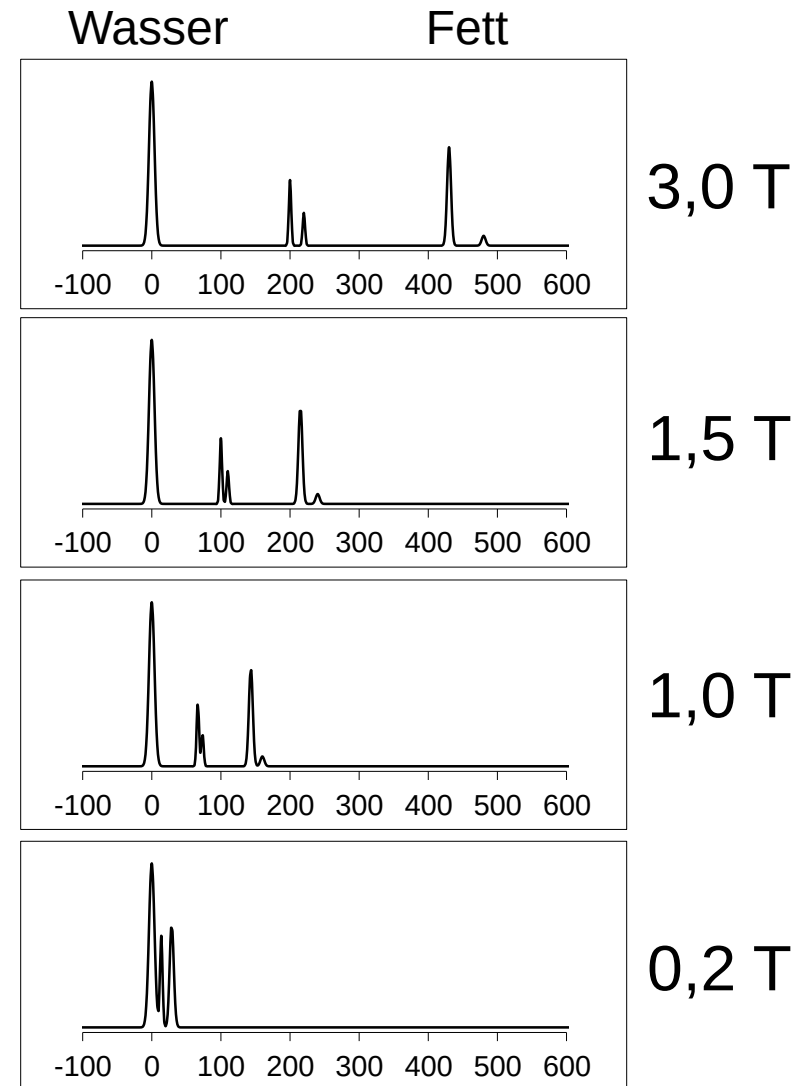
- Signal intensity is recorded as a function of frequency
- Signal stems from a single volume element (single-voxel-spectroscopy, SVA) or a small matrix (2D or 3D) – magnetic resonance spectroscopic imaging (MRSI) (16^2)
- Essential precondition: homogenous magnetic field for separation of signals
- no gradient during read-out!

MRS basics

At which field strengths can we do MRS?

Principally MRS is possible at all field strengths, but:

- Chemical shift (in Hz) depends on field strength
- At low field strength, separation of the signals is not possible
- in-vivo MRS at B_0 of 1.5 T or higher
- in-vitro: spectrometers with B_0 up to 20T are applied

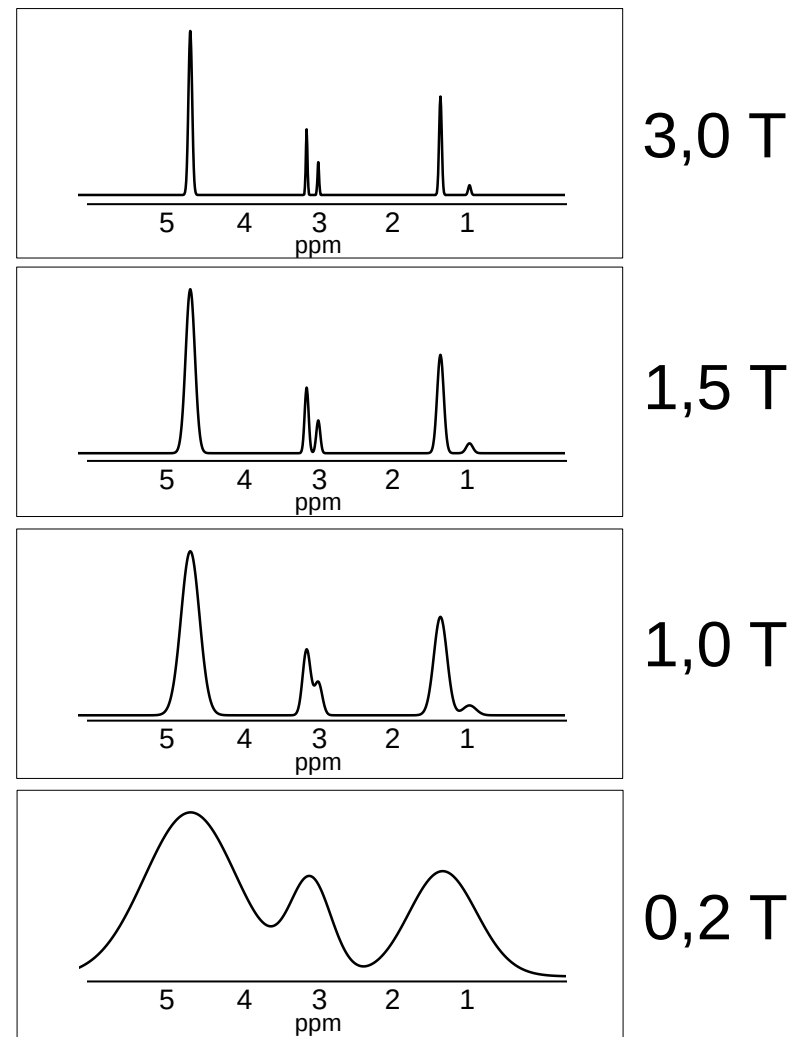


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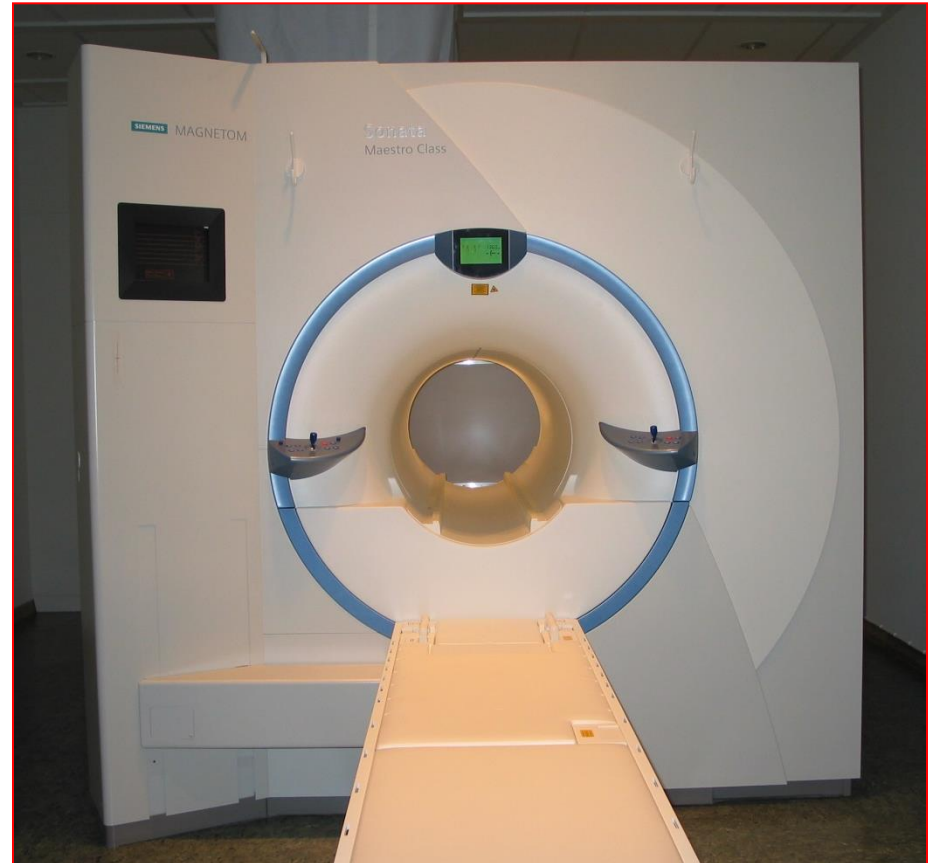


MRS basics

In-vitro vs. in-vivo



Spectrometer
(800 MHz, ca. 19 T)
Bruker Biospin



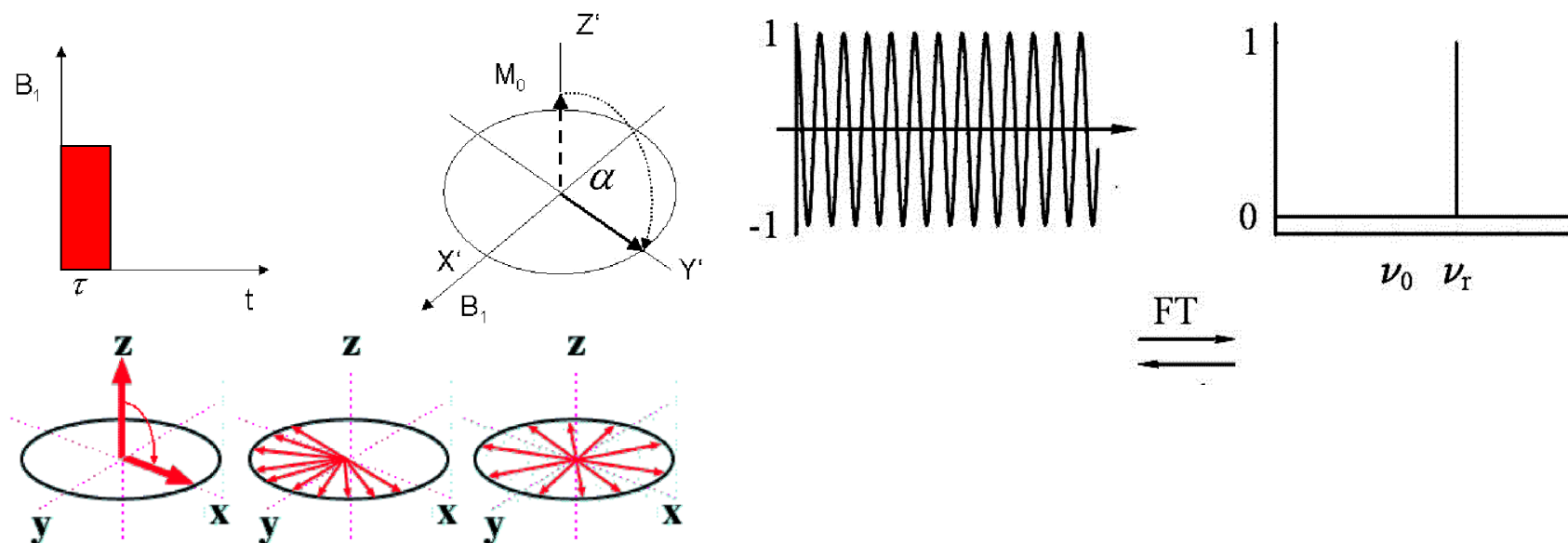
Whole body imager
(1,5 T Magnetom Sonata)
Siemens Healthcare

MRS basics

FT-Spectroscopy

FID-Sequence – One-Pulse-experiment

Irradiation of a 90°-pulse, read-out of signal



$$M_x(t) = M_0 \sin(\omega t + \varphi) e^{-(t/T_2^*)}$$

$$M_y(t) = M_0 \cos(\omega t + \varphi) e^{-(t/T_2^*)}$$

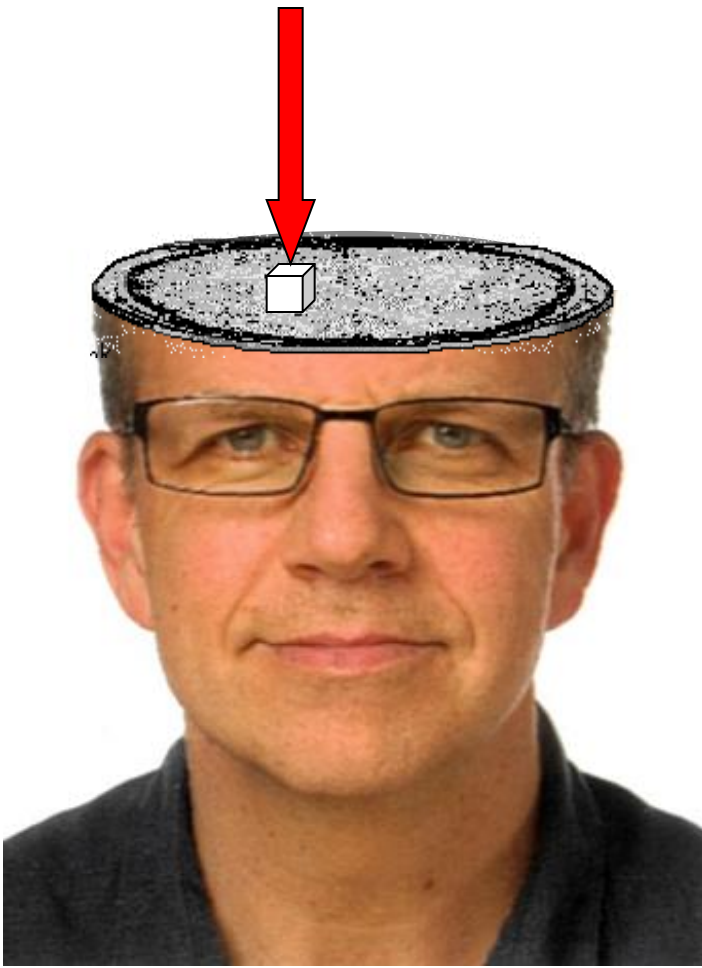
No volume selection!!!

Single-Voxel-Spectroscopy

Techniques for localization

Single-voxel-Spektroskopie (SVS)

VOI

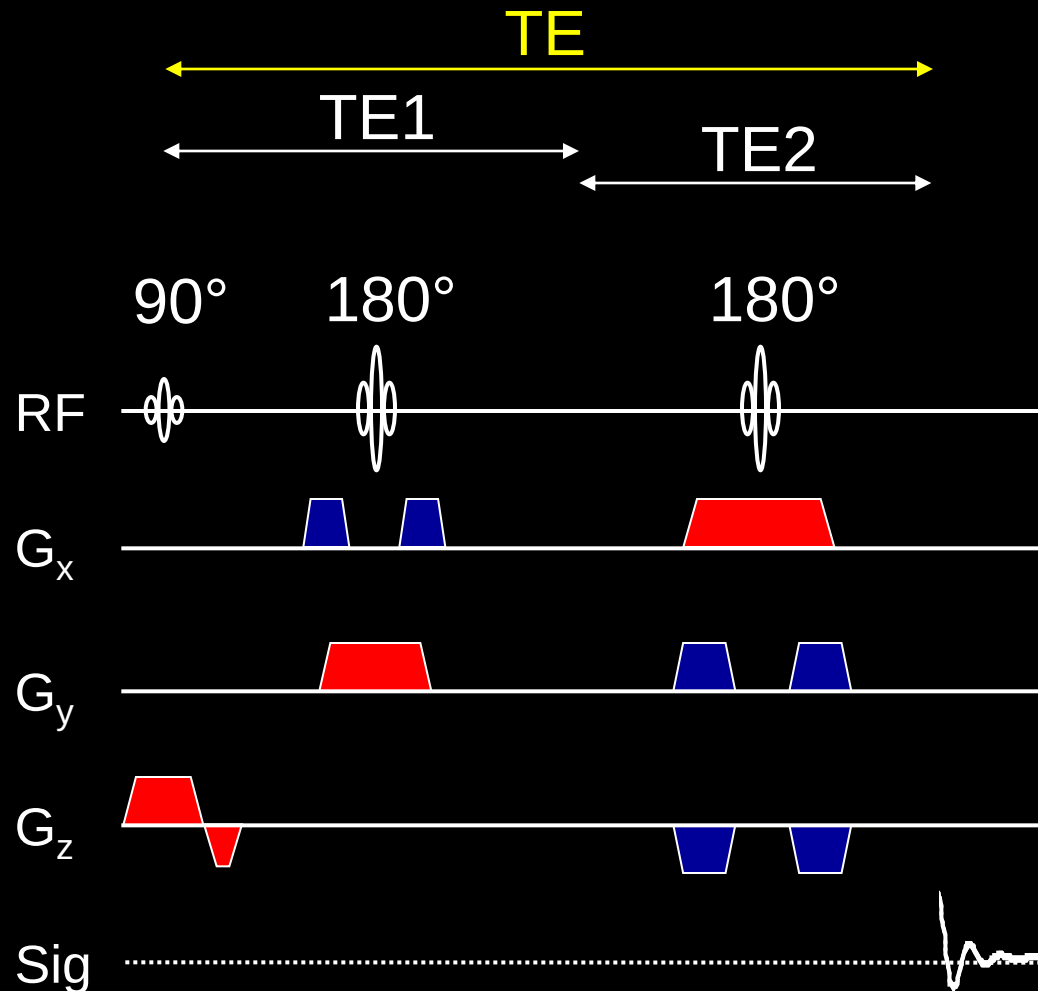


Signal should be recorded from a small volume:

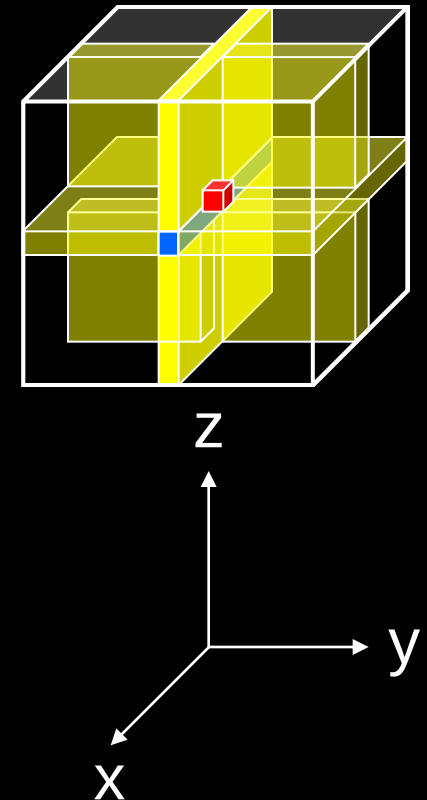
→ How to select a „volume of interest“ (VOI)?

Single-Voxel-Spectroscopy

Point-RESolved-Spectroscopy (PRESS)

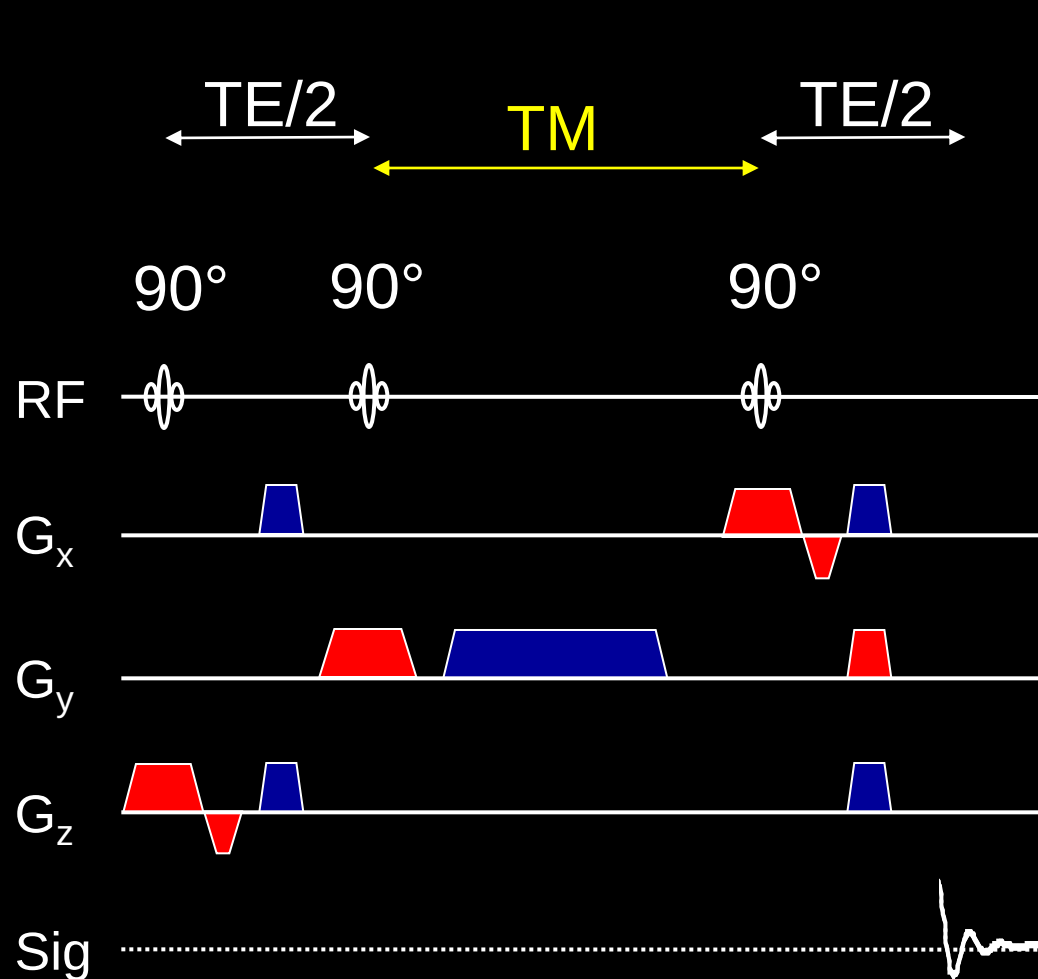


 Slice selection
 Spoiler or Crusher



Single-Voxel-Spectroscopy

STimulated Echo Acquisition Mode (STEAM)



 Slice selection
 Spoiler or Crusher

