MR: From Past to Future

MR'da Geçmişten Bugüne

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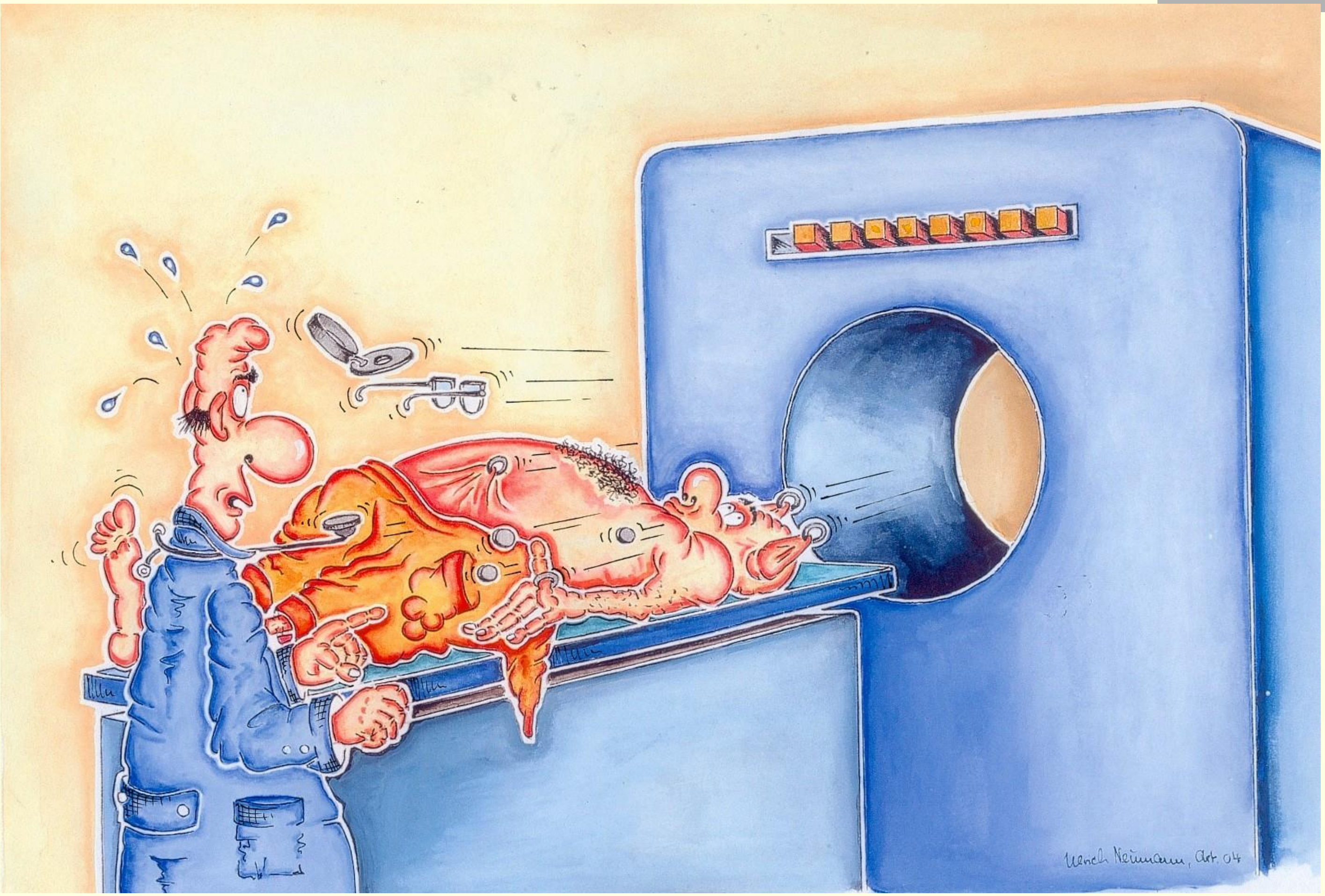
Tübingen‘s Hölderlin tower at the river ‚Neckar‘

Stereotactic RT treatment

The very beginnig:
What is magnetism?

Historic origin of magnetic mineral:

Magnetic forces of MR scanners



U. Neumann, Germany

Magnetic forces of MR units

Siemens

MR safety video clip with sound!

Nuclear Magnetic Resonance (NMR)

- NMR experiments by I. Rabi (1939 nuclear spin),
  F. Bloch/E. Purcell (1945 NMR discovery)
  R. Ernst (1975 Fourier-Spectroscopy)
- Images from MR:
  - P. Lauterbur: Space encoding by gradient fields (1973)
  - P. Mansfield: Image reconstruction,
    Echo-Planar-Imaging (EPI) (1977)
- What about the third: R. Damadian (1971,1977) ?
  Identifying tumor by NMR from its relaxation time !??

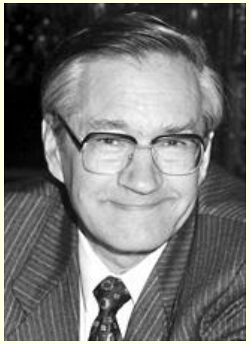
History of MR related Nobel prizes

- Physics: 1944: I. Rabi

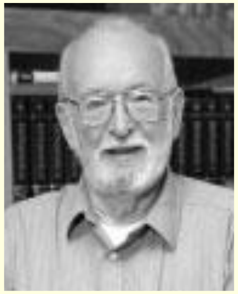
  1952: F. Bloch and E. Purcell



- Chemistry: 1991: R. Ernst,



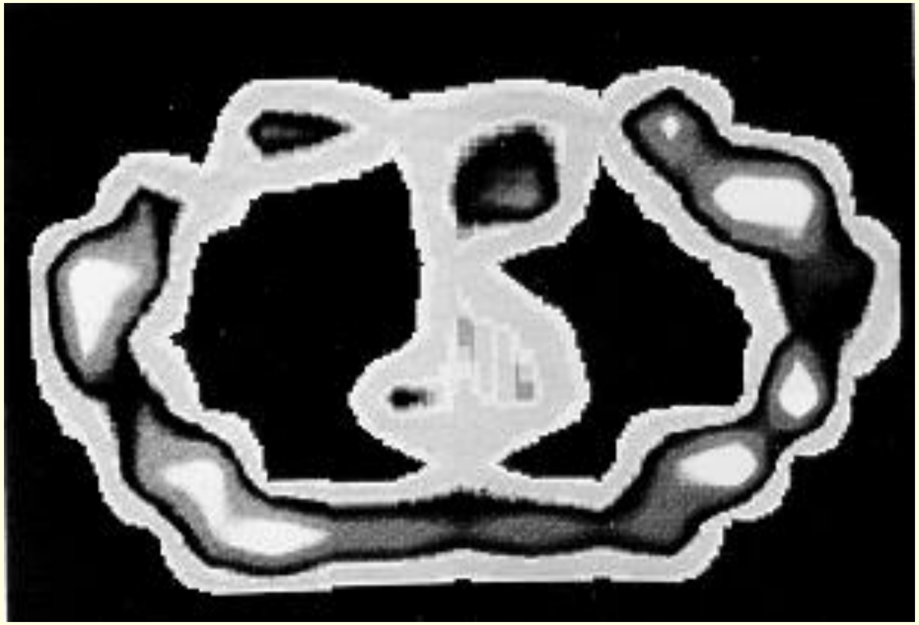
- Medicine: 2003: P. Mansfield and P. Lauterbur

History of MR Imaging: Technology of 1977

MR Scanner: „Indomitable“

R. Damadian



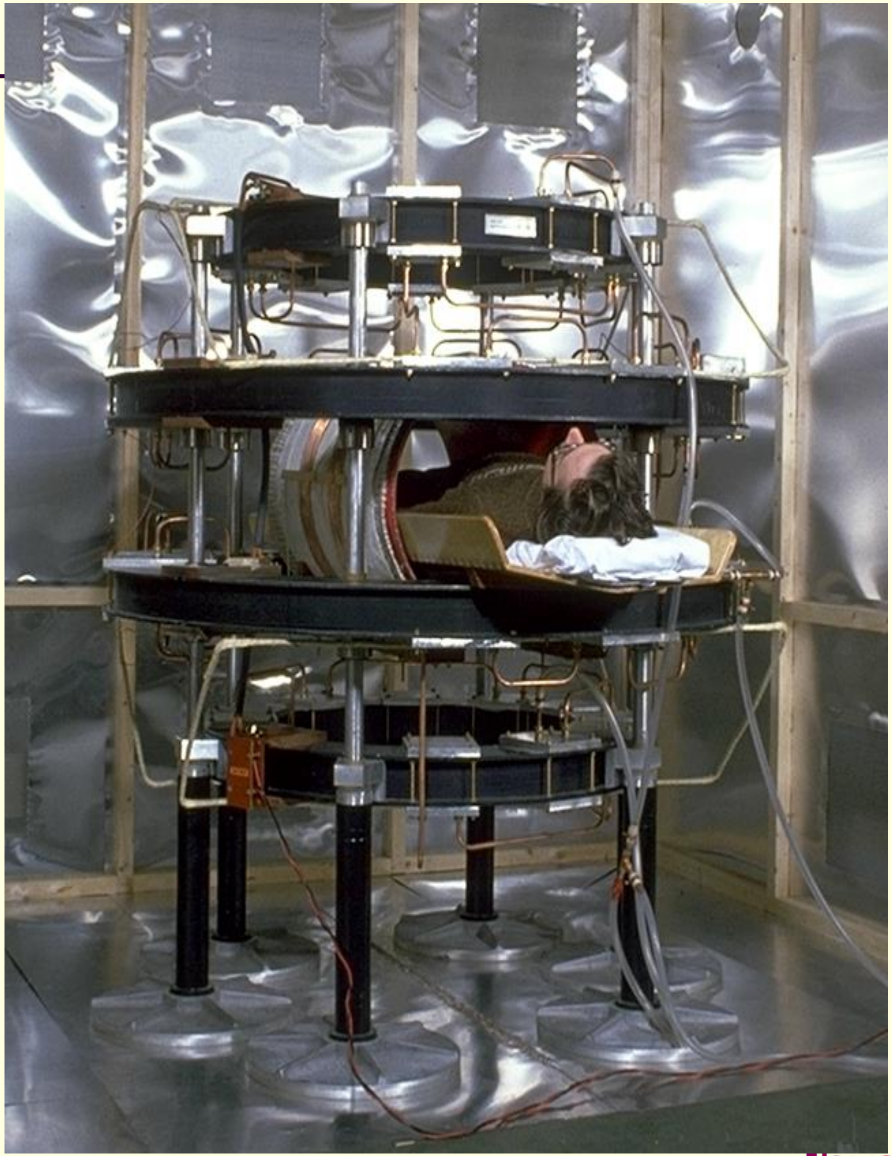
First MR image of Larry Minkoff (took 1h to scan!)

MR scanner in Aberdeen, UK
“Mark I”, 1979

John Mallard:

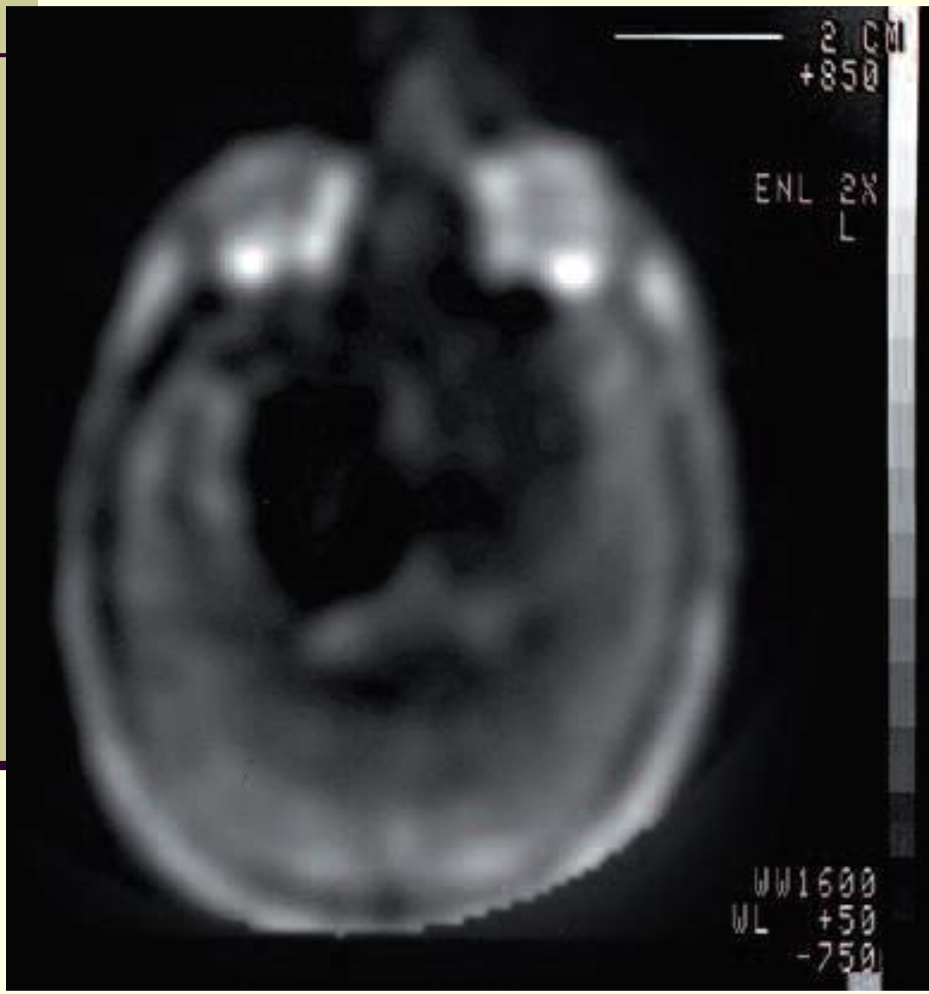
Spin-wrap technique: Gradient and phase encoding for 2D image slice

- World’s first clinically-used MRI scanner
- 0.04 Tesla (1.7 MHz)
- Used to image 1000 patients in early 1980s



Courtesy P. Sharp, Aberdeen

First image of human brain in Europe with a Philips MR (0.15T)

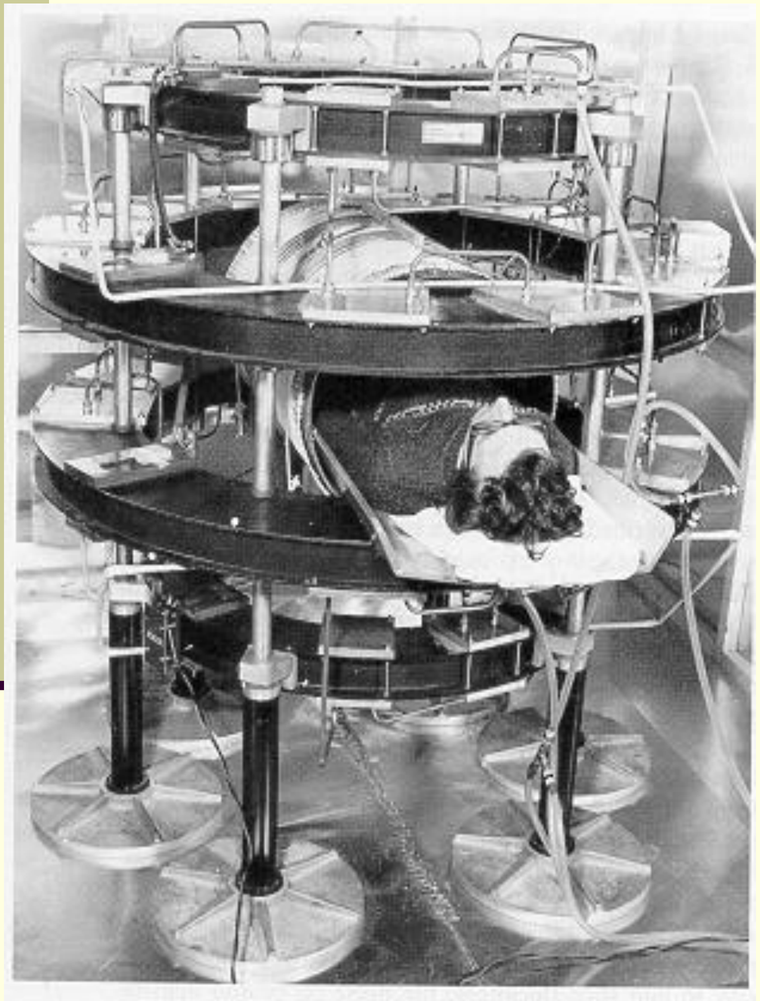


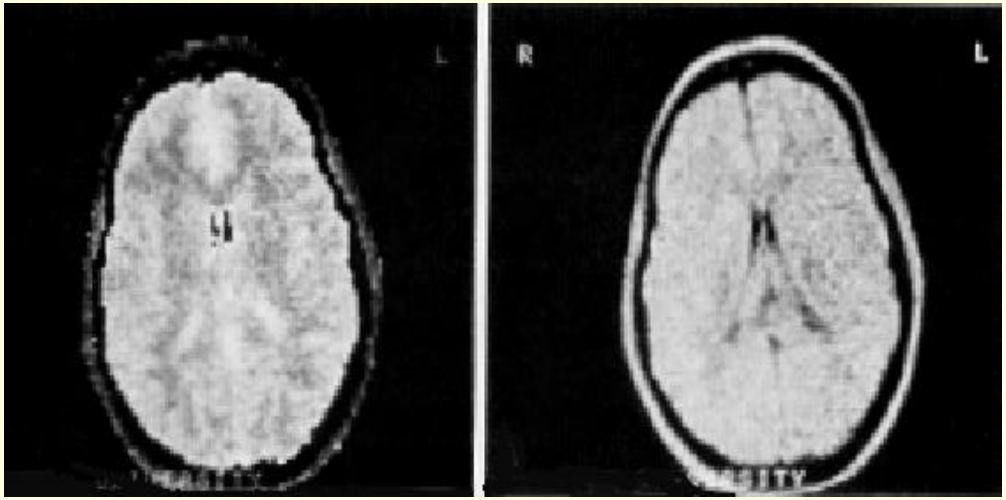
December 3, 1980:
½ h for acquisition of single slice, several hours for reconstruction.

Image: head of André Luiten, standing at the MR scanner above

 From: MedicaMundi 46/1 April 2002

MR scanner in Aberdeen, UK
"Mark I" MRI, 1981





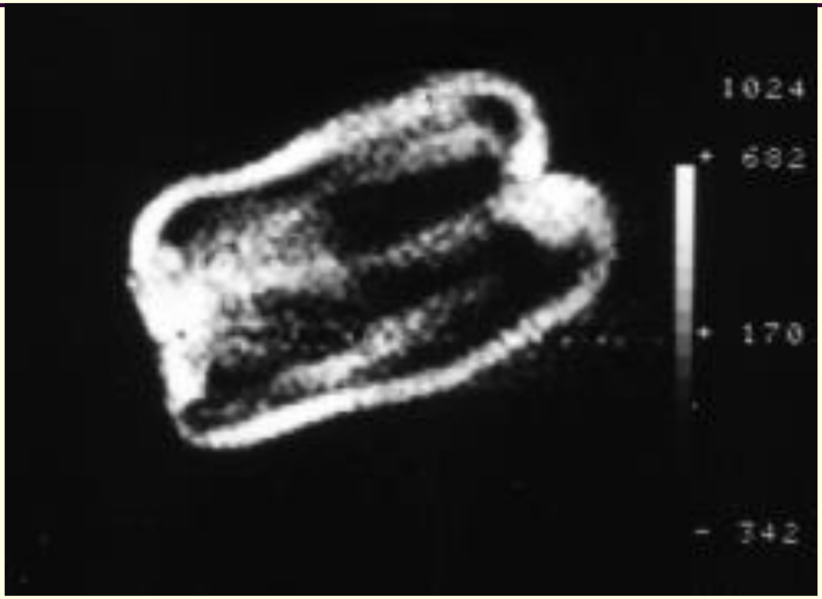
MR-images 1981
Acquisition time ~45 min.

Aberdeen “Mark II” MRI, 1983

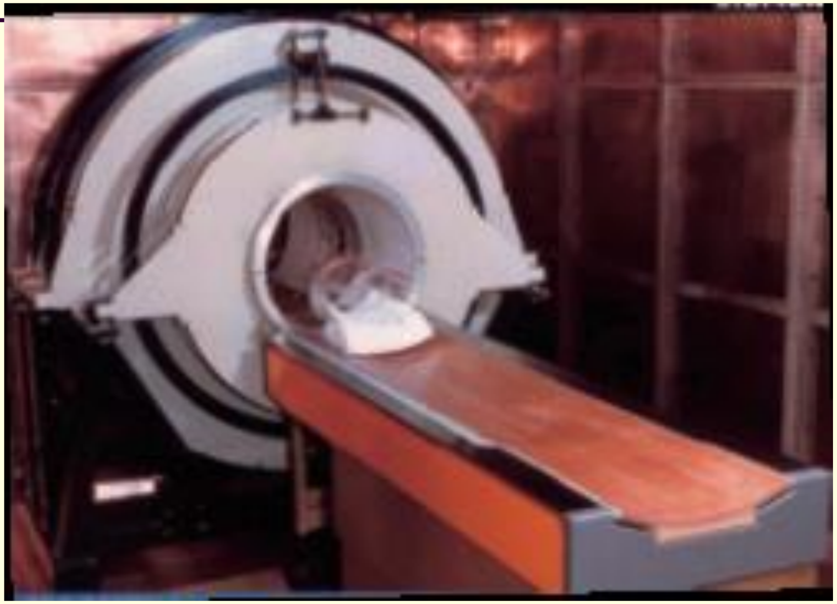
- Same basic design as Mark I
- 0.08 Tesla (3.4 MHz)
- Used to image 5000 patients between 1983 and 1992
- Formed basis of commercial scanners (M&D Technology and Asahi Medical)

Courtesy P. Sharp, Aberdeen

Early MRI at Siemens/Germany



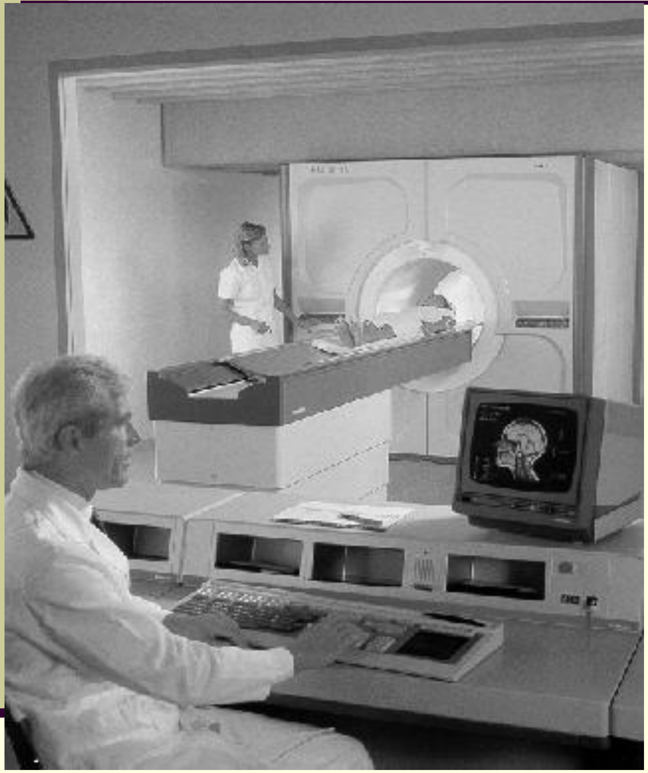
First MR-image 1979:
a pepper

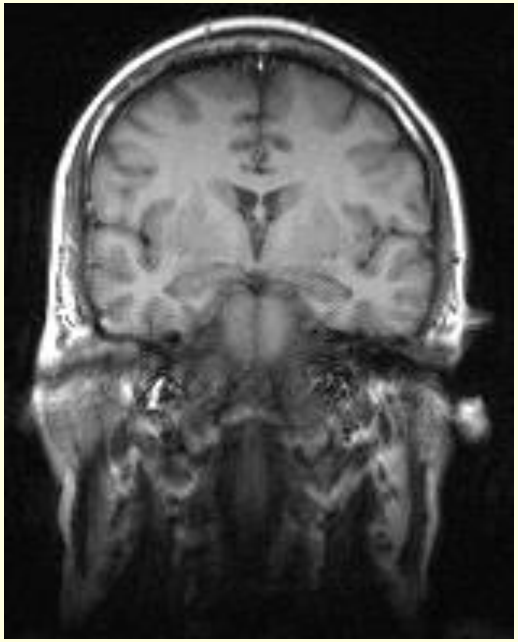


Resistive magnet,
field strength: 0.2 Tesla

Courtesy Siemens

Siemens MR technology in 1985



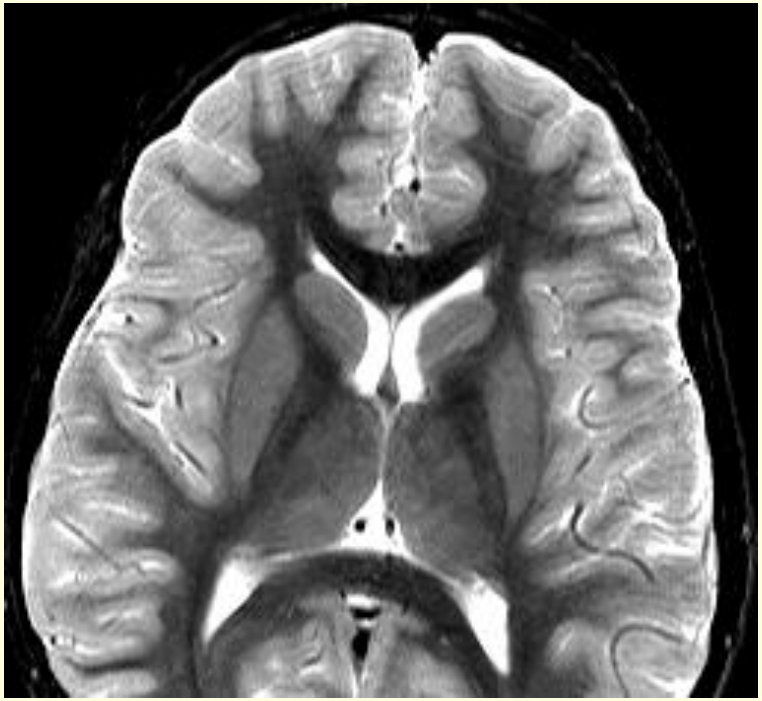


Siemens Magnetom, 1.5 T,
Gradient field: 3 mT/m

First MR in Tübingen:
Acquisition time ~10 min.

Courtesy F. Schick, Tübingen

Siemens MR scanner at Tübingen



1.5 Tesla, Gradient field: 40 mT/m

T_2w-Turbo-SE-tomogram
Acquisition time: ~1 min.

Courtesy P. Sharp, Aberdeen

MRI scanners for special purposes: Orthopedics (ESAOTE/Italy)

Extremety-MRI system C-Scan, ESAOTE

MRI scanners for special purposes: Veterenarian orthopedics

(ESAOTE/Italy)

MRI scanners for special purposes:

Upright patient scanning

FONAR pMRI 0.6 T, resistive magnet

(Field fOcussed Nuclear mAgnetic Resonance, R. Damadian's company)

 Courtesy P. Sharp, Aberdeen

Open MR: Philips Panorama HFO

1 Tesla, superconducting magnet („donut“),
MR simulation for radiotherapy possible

Courtesy Philips

Now comes the physics!

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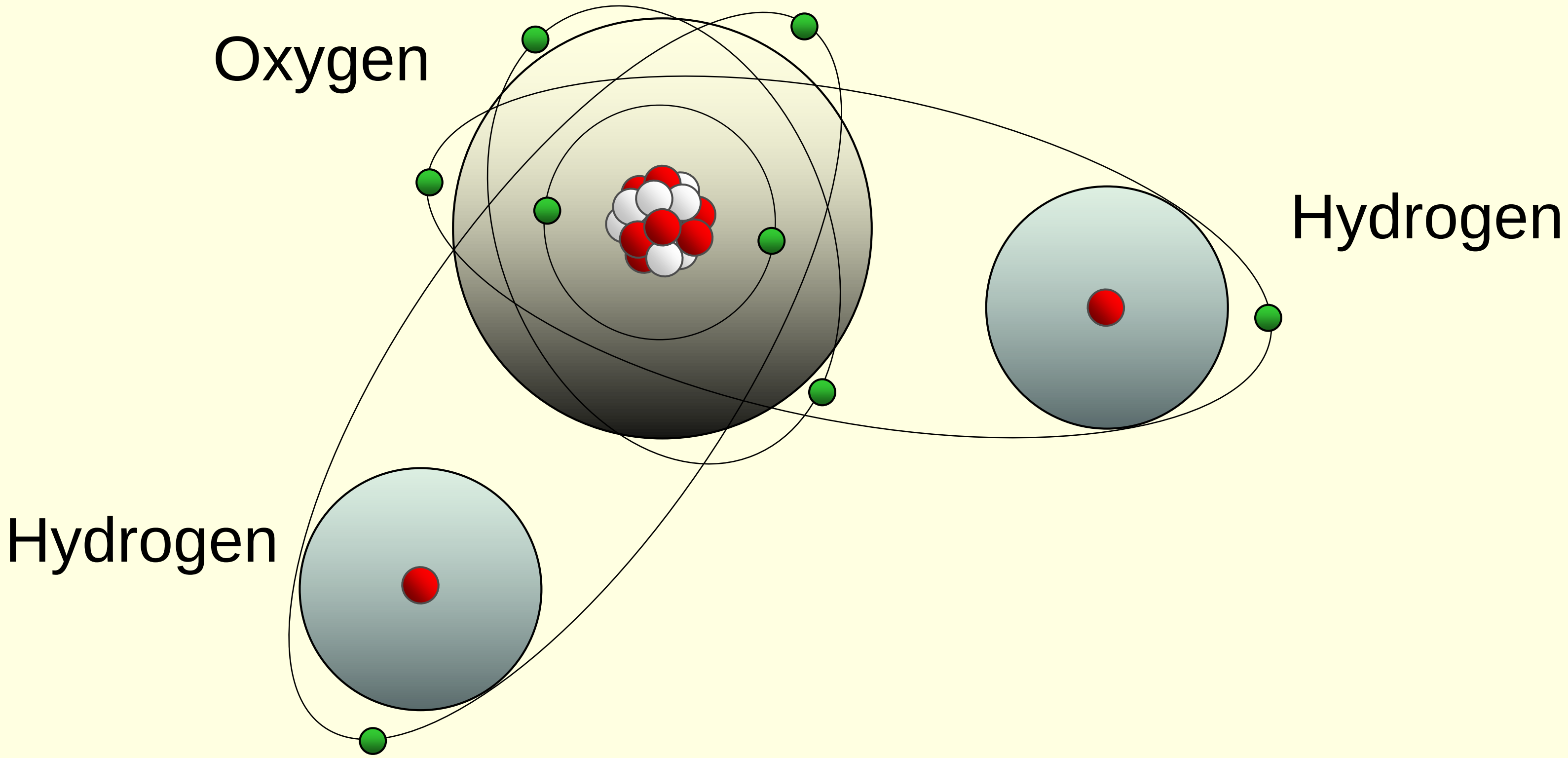
Nuclear Magnetism

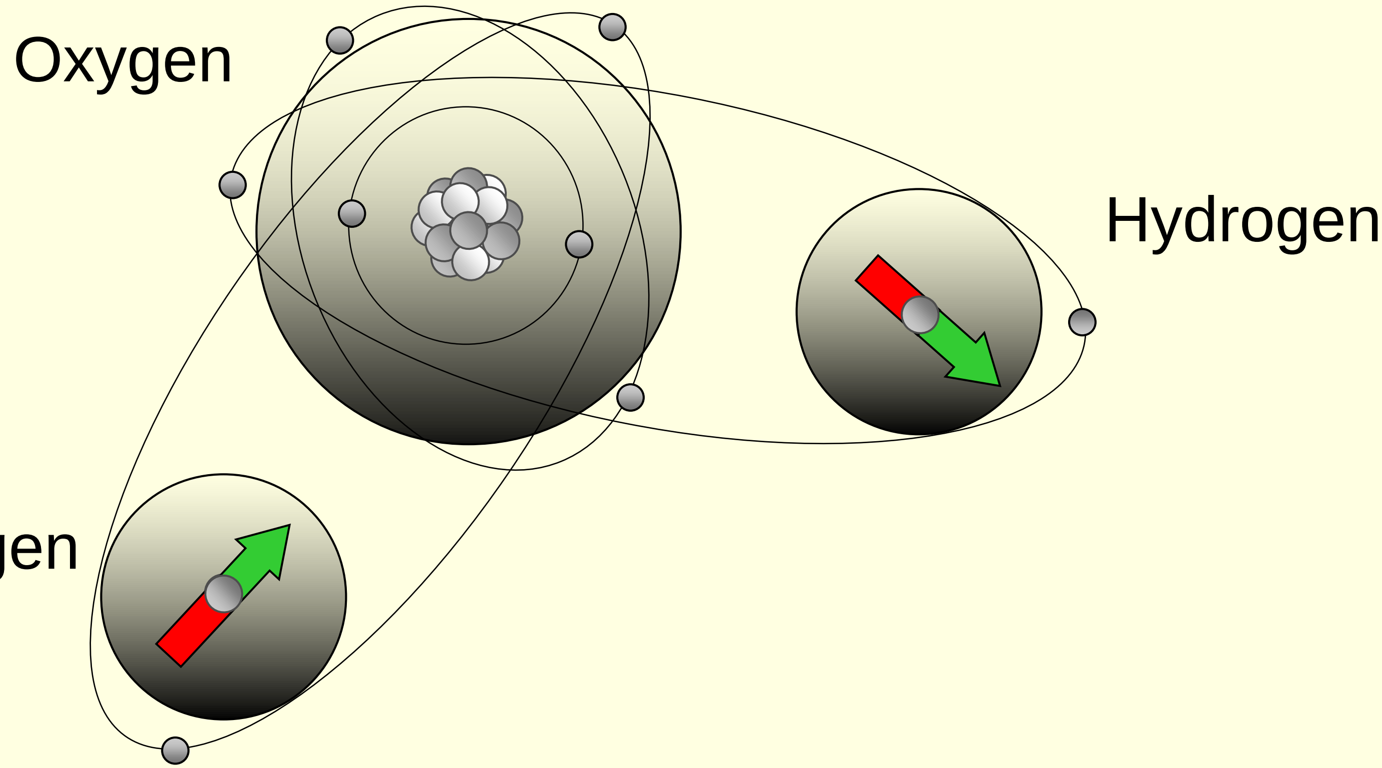
The water molecule:



Courtesy: Klaas Prüssmann, ETH Zürich

Nuclear Magnetism

The water molecule:

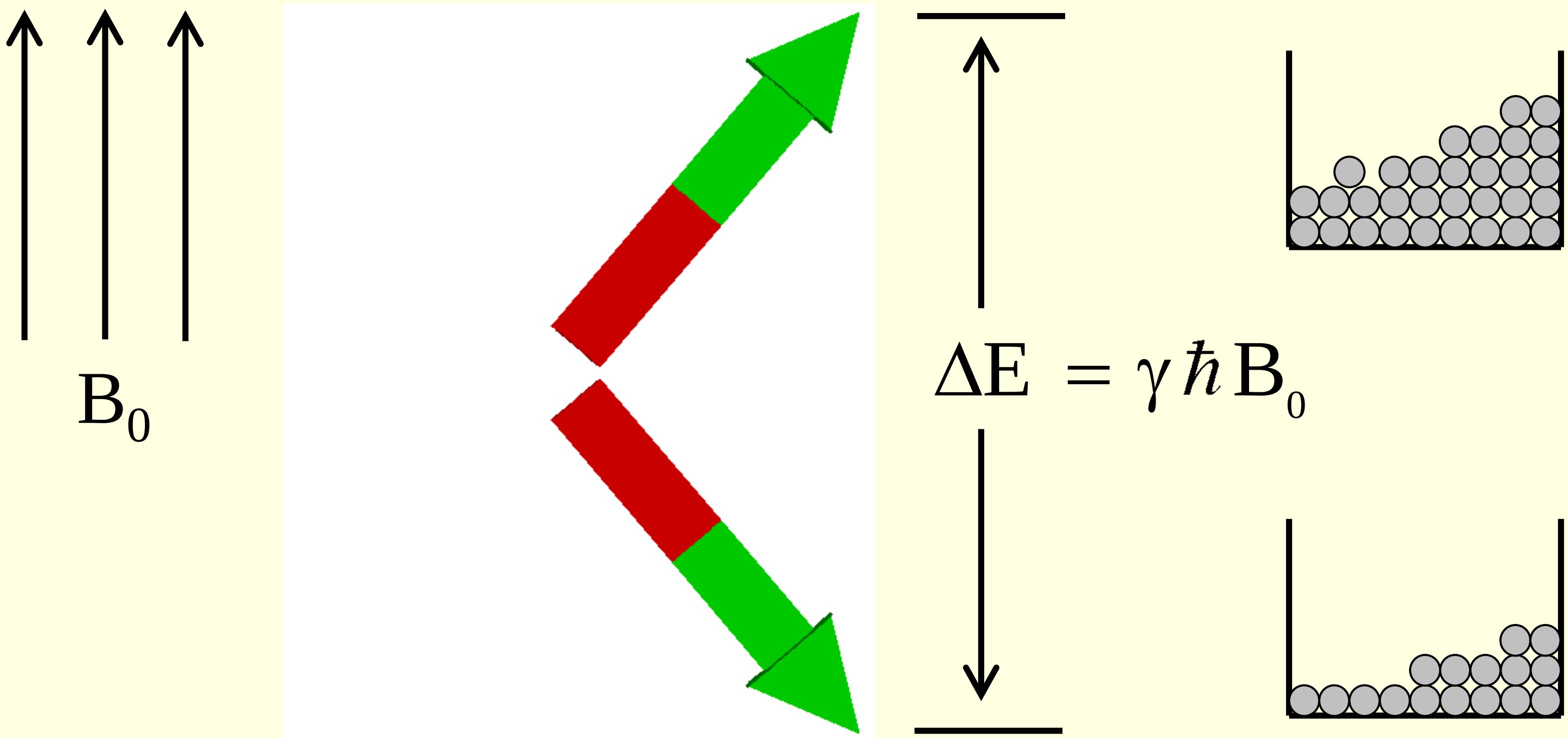


Courtesy: Klaas Prüssmann, ETH Zürich

Nuclear Magnetism

^{1}H nuclei in a magnetic field

<u>Quantisation</u>



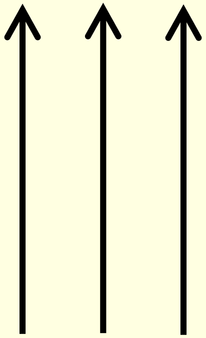
$$\Delta E = \gamma \hbar B_0$$

(rel. spin excess ≈ 3.3 10^{-6} per Tesla at RT)

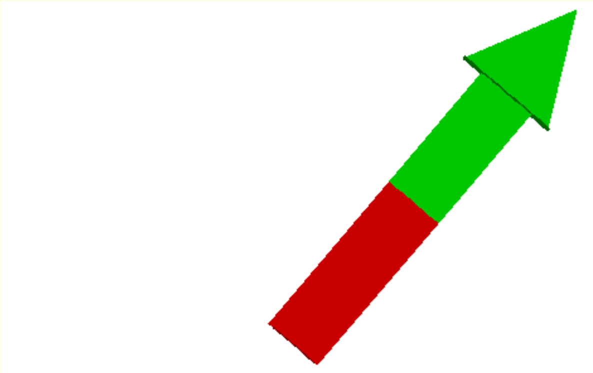
Courtesy: Klaas Prüssmann, ETH Zürich

Nuclear Magnetism

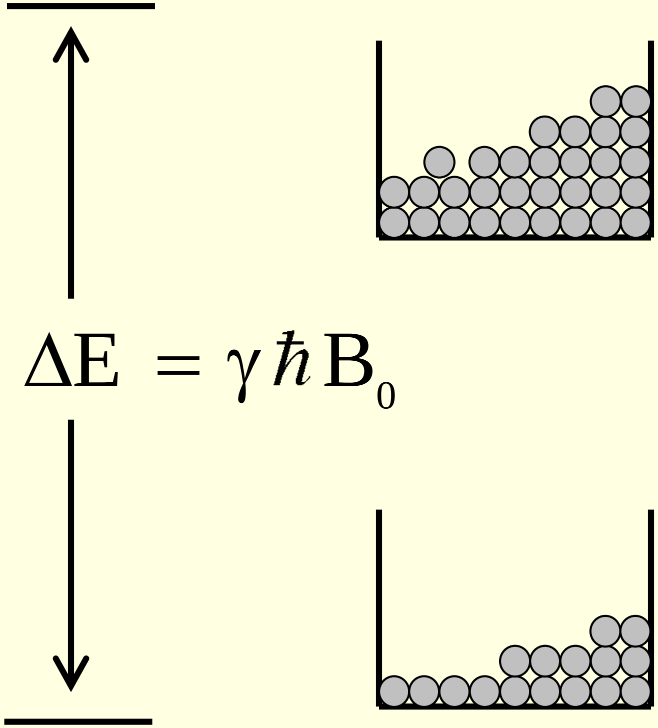
^{1}H nuclei in a magnetic field



B_0



Quantisation



$$\Delta E = \gamma \hbar B_0$$

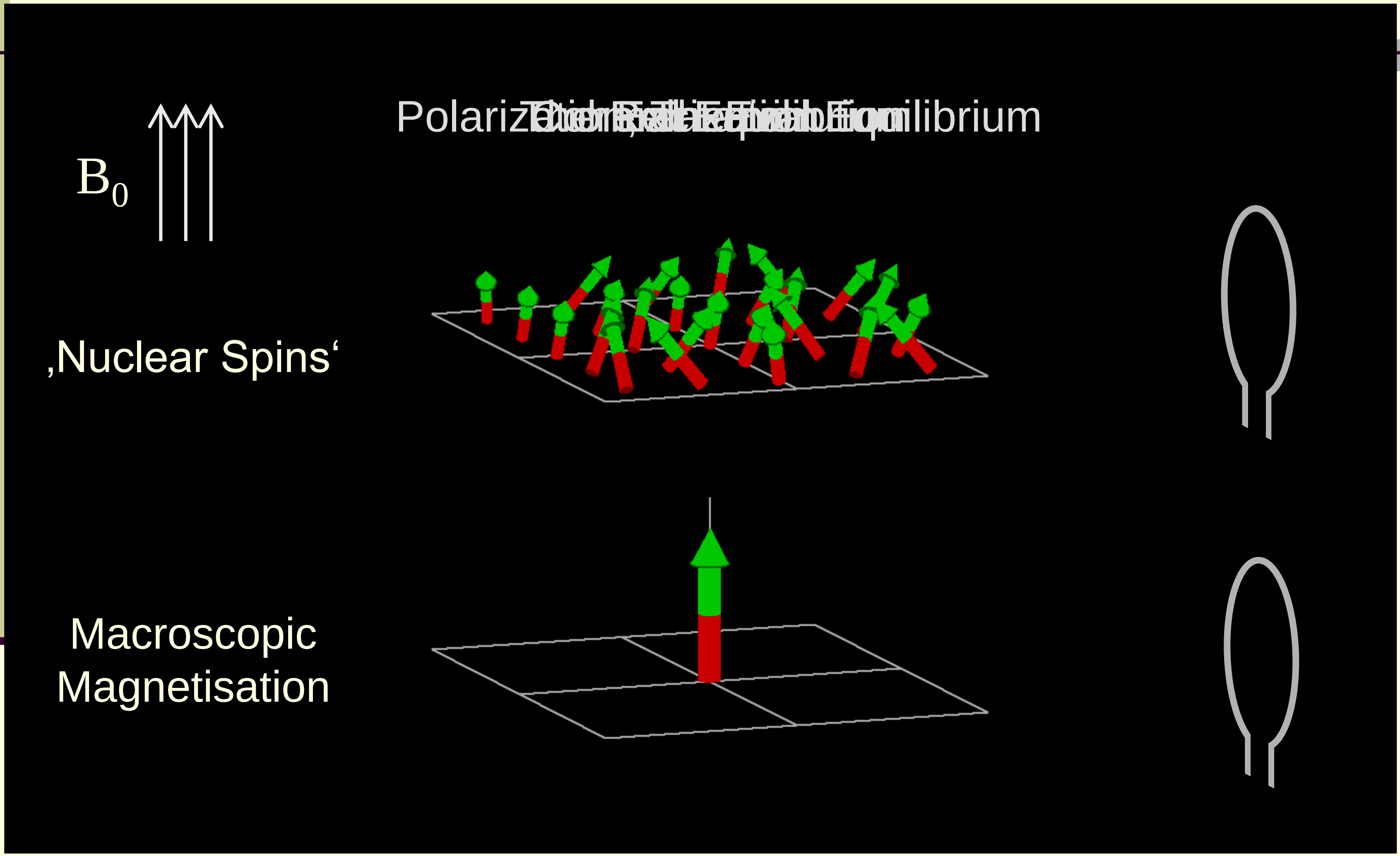
Precession: $\omega_L = \gamma B_0$

⇨ 42 MHz at 1 Tesla

(rel. spin excess ≈ 3.3 10^{-6} per Tesla at RT)

Courtesy: Klaas Prüssmann, ETH Zürich

Physics behind MR images



 Courtesy: Klaas Prüssmann, ETH Zürich

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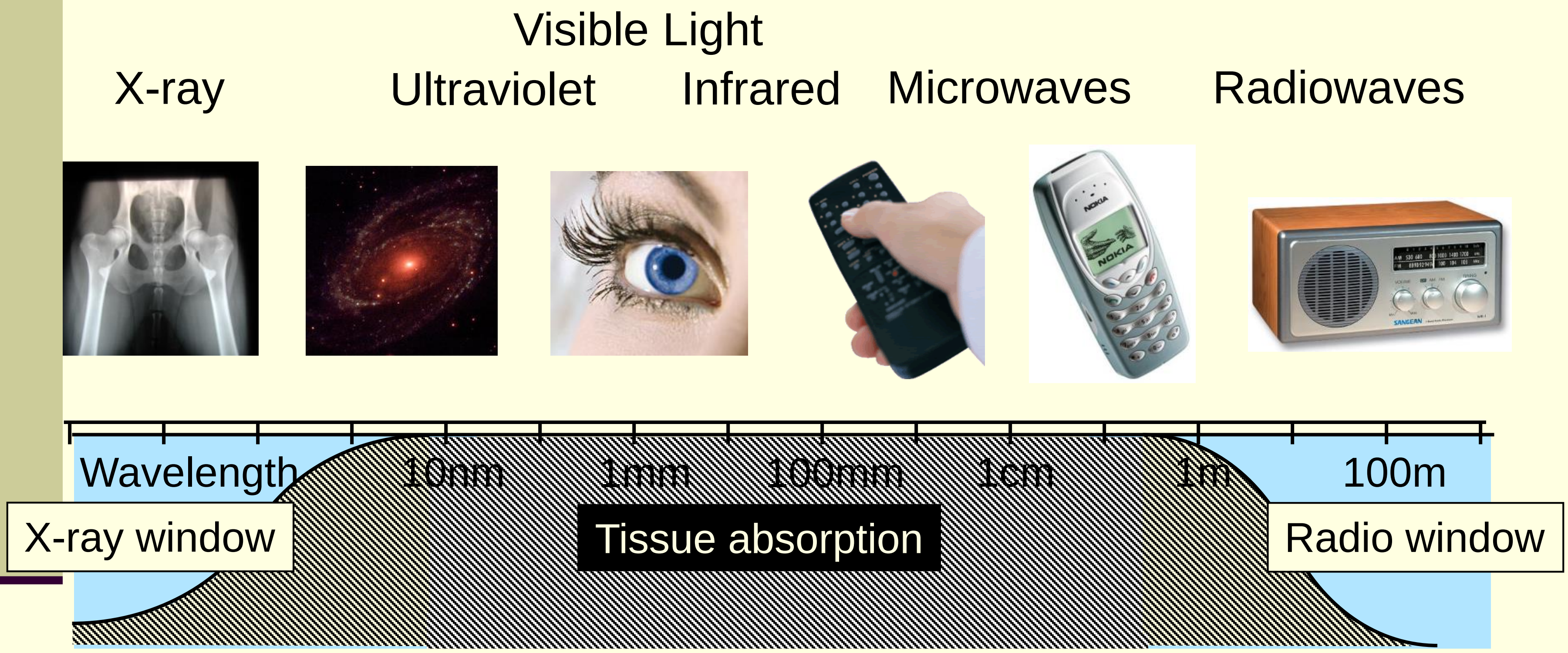
Physics behind MR images



Courtesy: Klaas Prüssmann, ETH Zürich

Physics behind MR images

Resolving resonant magnetization:

- Focussing ?
- Collimation ?

⇒ Diffraction limit $\Delta x \geq \frac{\lambda}{2}$

The radio-frequency window, on the other hand, was not exploited until 1972. This is not astonishing considering the achievable resolution, which is usually limited by the wavelength of the applied radiation through the uncertainty relation. The maximum radio-frequency useful for imaging is about 100 MHz, leading to a resolution of 3 m which is not sufficient even for imaging elephants.

The crucial idea, as first proposed by Lauterbur (10.1, 10.2), is to utilize a magnetic field *gradient* to disperse the NMR resonance fre-

R.R. Ernst et al., Principles of NMR in One and Two Dimensions (1987)

 Courtesy: Klaas Prüssmann, ETH Zürich

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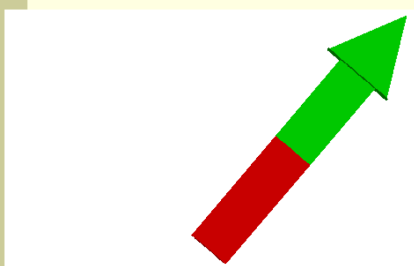
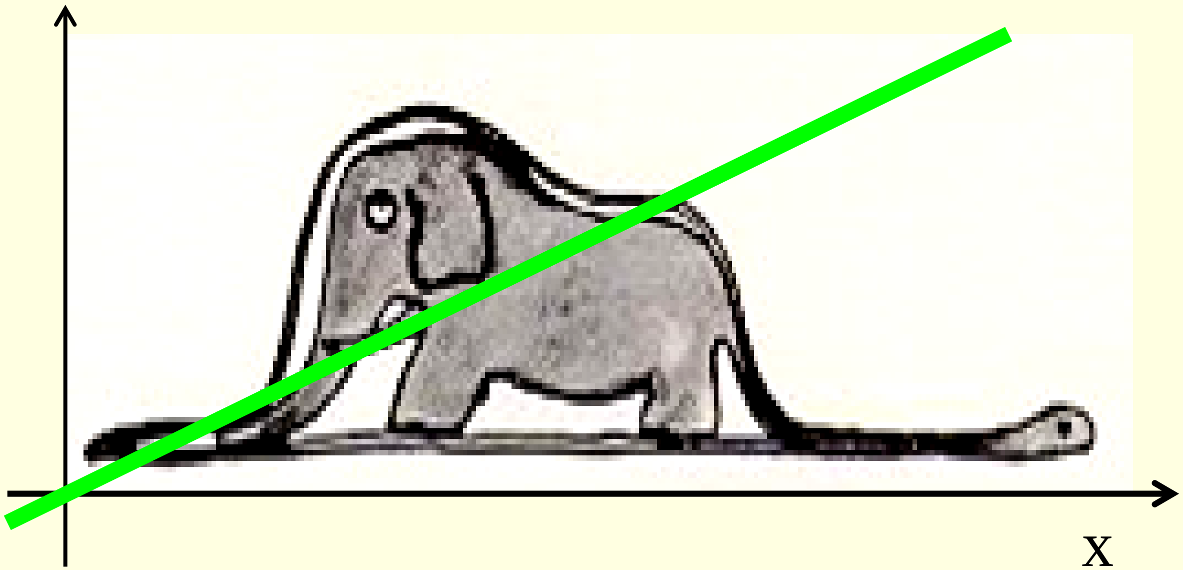
Physics behind MR images

Superimpose magnetic gradient field

$$\omega_L(x) = \gamma\,(B_0 + G\,x)$$





Encodes spin distribution in resonance spectrum

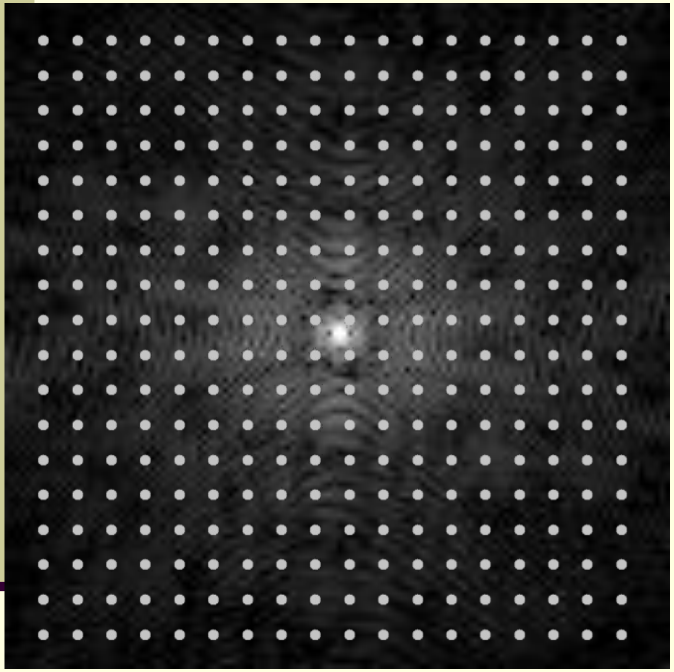
⇒ MR Imaging = Spectroscopy

P. C. Lauterbur (1973)

Courtesy: Klaas Prüssmann, ETH Zürich

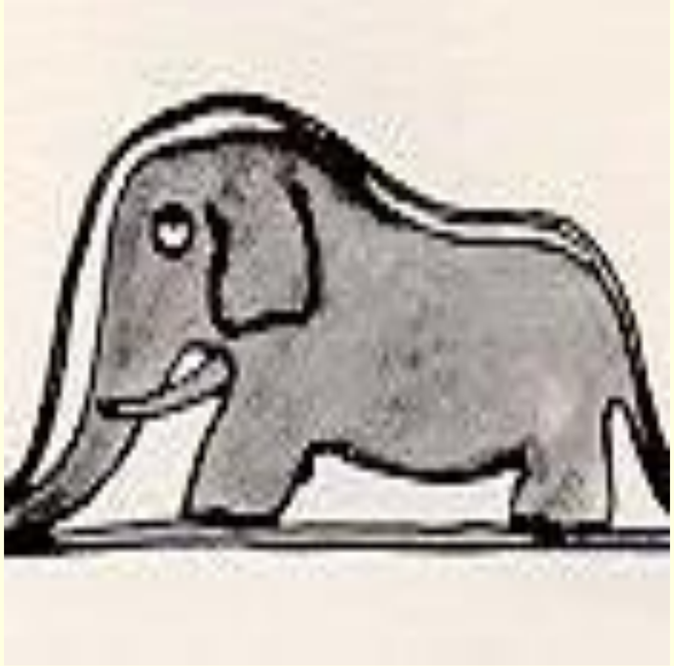
Physics behind MR images

Fourier Space



Fourier Transform

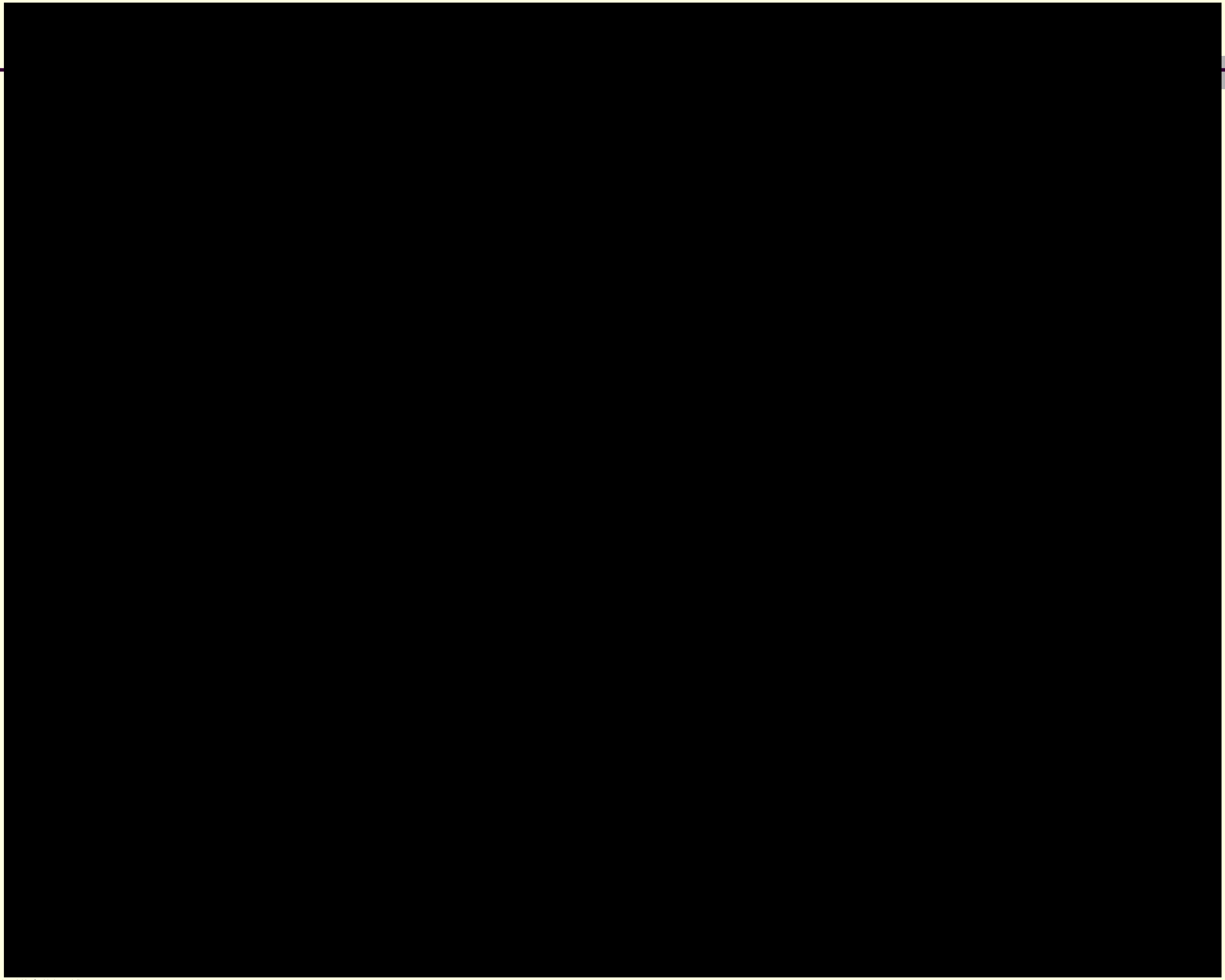
Image Space



Kumar, Welti, Ernst (1975)

Courtesy: Klaas Prüssmann, ETH Zürich

Let' s repeat the physics behind MR:



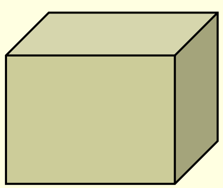
Siemens MR physics video clip with sound!

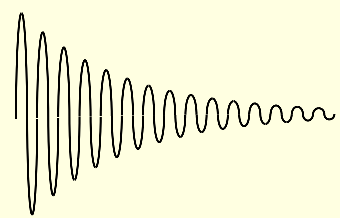
More information from the MR signal !

RF Transmission	*Spin Evolution*	*Acquisition*
Maxwell	Bloch	Maxwell

Spatial Encoding: $\omega_L(\vec{r}) = \omega_0 + \gamma \vec{G} \cdot \vec{r}$

Position encoded in: Precession frequency, phase — Signal frequency, phase





2(3) spatial dimensions → 1 temporal dimension

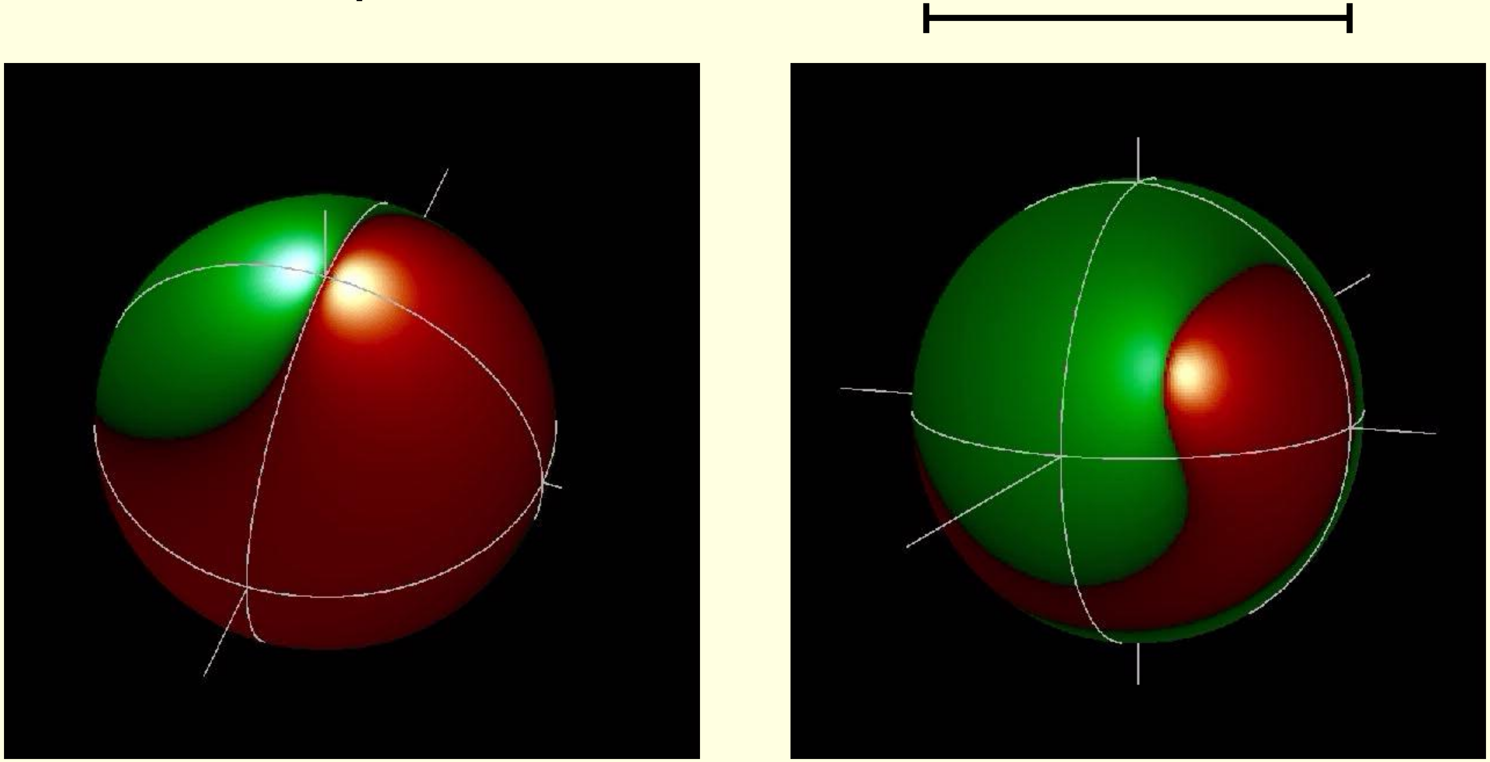
Courtesy: Klaas Prüssmann, ETH Zürich

More information from the MR signal !

Precessing magnetic moment in a sphere

$\varepsilon_r = 80, \ \omega/2\pi = 64$ MHz

25 cm



Flux through surface: Positive Negative

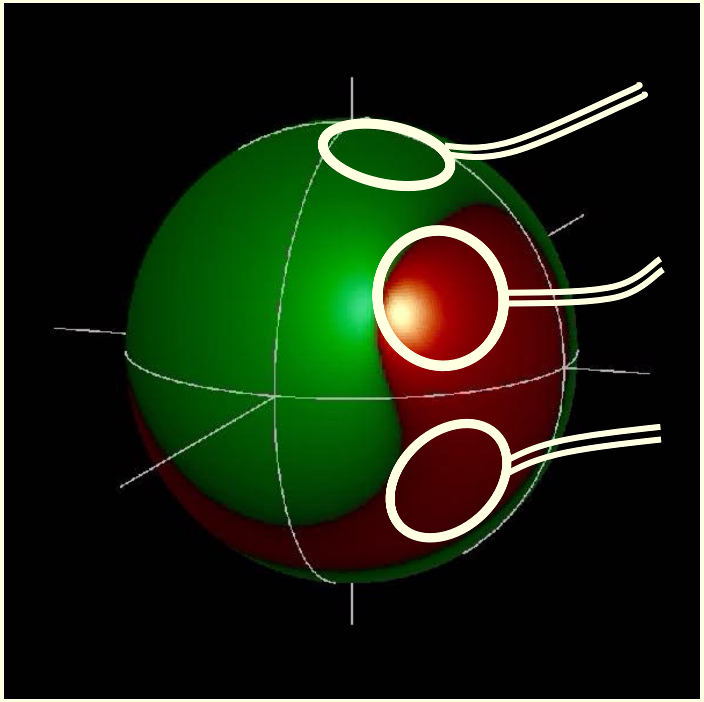
Spatial degrees of freedom: View from different positions

Courtesy: Klaas Prüssmann, ETH Zürich

More information from the MR signal !

$\varepsilon_r = 80$, $\omega/2\pi = 64$ MHz

25 cm



Position of magnetic moment is encoded in flux pattern!

How capture this information ?

Sample with multiple detectors !

Coil sensitivity $$s_i(\vec{r}) = \int_{\text{coil } i} \vec{\Phi} \cdot d\vec{n}$$

Flux through surface: Positive Negative

 Courtesy: Klaas Prüssmann, ETH Zürich

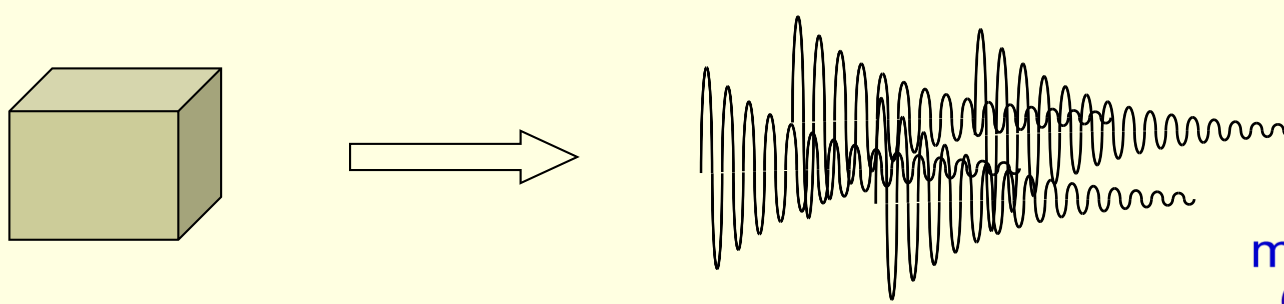
Phased array antennas

RF Transmission	*Spin Evolution*	*Acquisition*
Maxwell	Bloch	Maxwell

Spatial Encoding: $\omega_L(\vec{r}) = \omega_0 + \gamma \vec{G} \cdot \vec{r}$ $\quad s_i(\vec{r}) = \int \vec{\Phi}_{\vec{r}} \cdot d\vec{n}$

Position encoded in: Precession frequency, phase — Signal frequency, phase

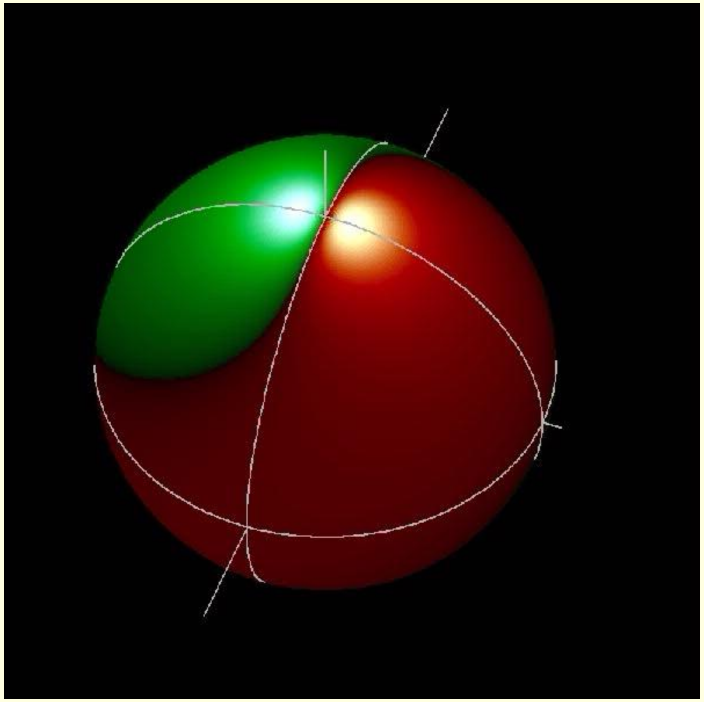
'Array dimension'



M. Buchgeister, Antalya 2007 Courtesy: Klaas Prüssmann, ETH Zürich

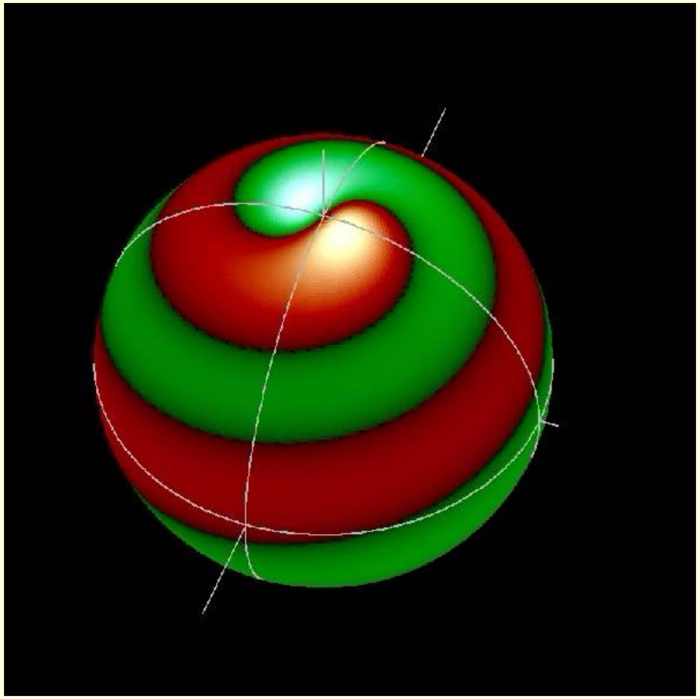
Limits of Parallel MRI

1.5 Tesla
f = 64 MHz



25 cm

7 Tesla
f = 300 MHz



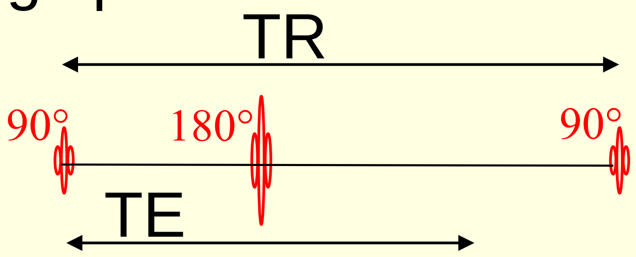
25 cm

Courtesy: Klaas Prüssmann, ETH Zürich

What causes the contrast in MR images?

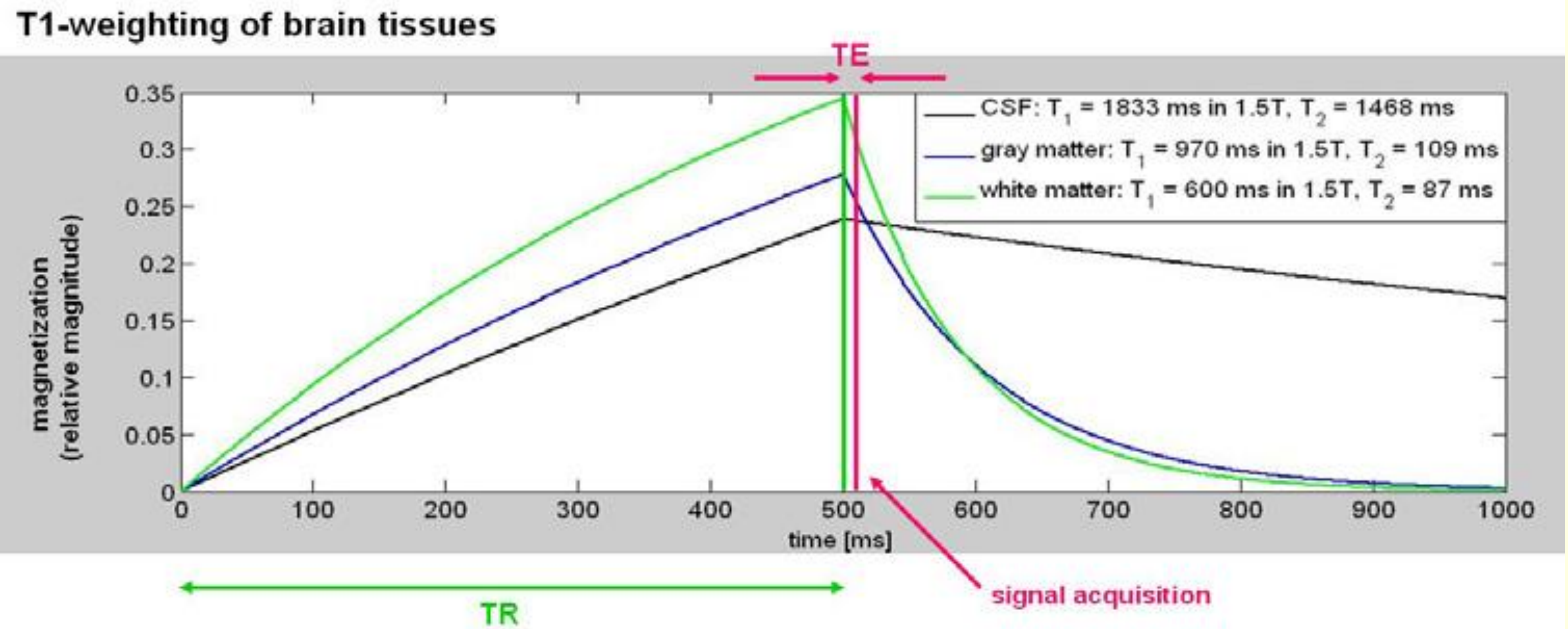
Relaxation time: T1

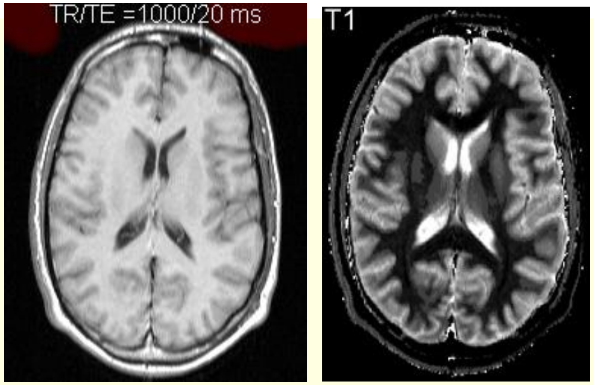
- Influence of spin – lattice coupling („transfer of heat“) within matter
- Longitudinal magnetisation:
  cannot be measured directly!
- Requires use of 90° RF puls every TR to flip longitudinal magnetization into transversal plane;
- a 180° RF puls every TE/2 reverses the spins and causes an echo of the dephasing spins
- Spin echo puls sequence:
  (TR: repetition, TE: echo time)



  - TR around T1, eg. ~1 s
  - TE short (shorter than T2), eg. ~20 ms

Relaxation time: T1

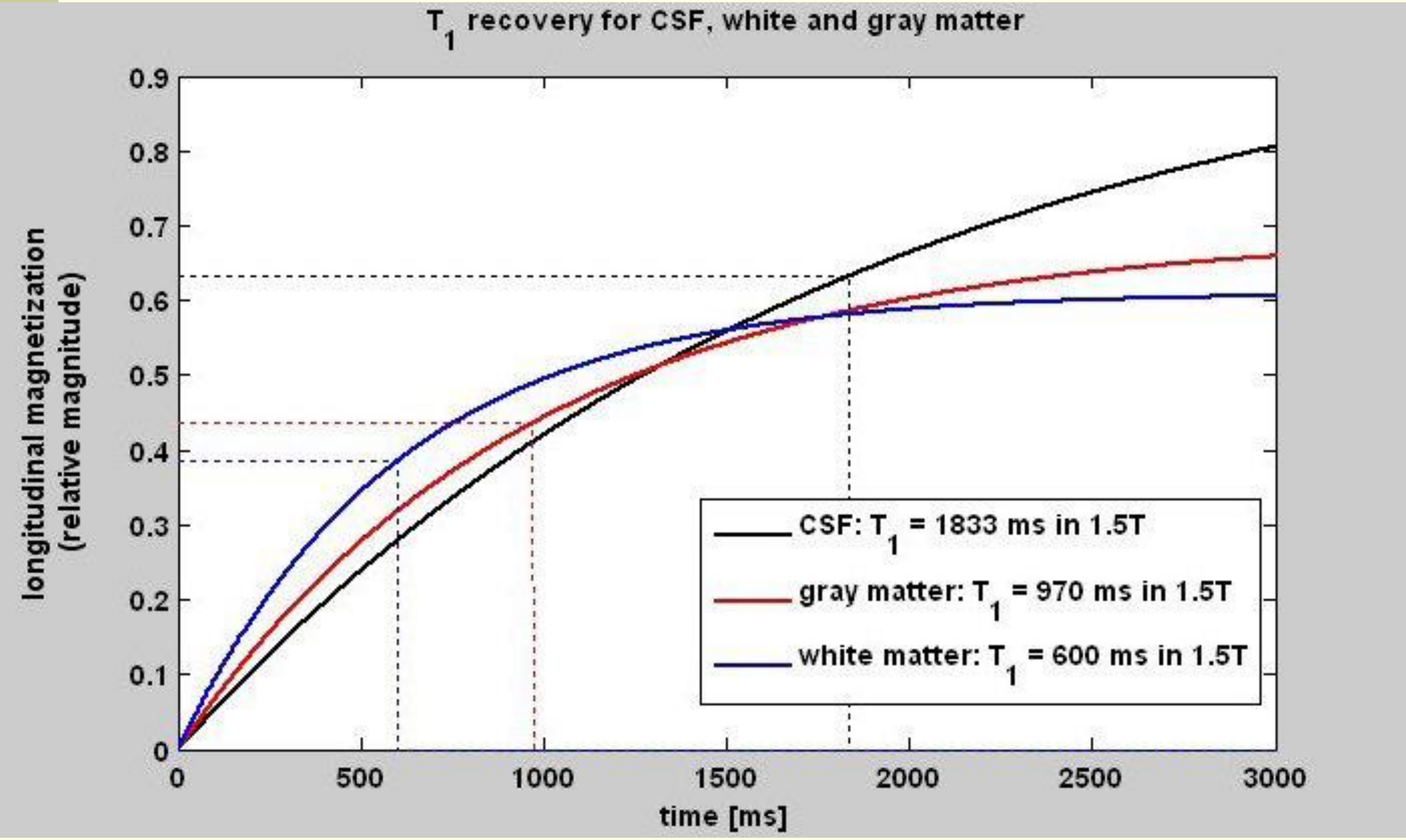




Left: T1-weighted image (left)
Right: image from the actual T1 values of the tissues.

From: Schlegel/Mahr, 3D Radiaton Therapy, 2007

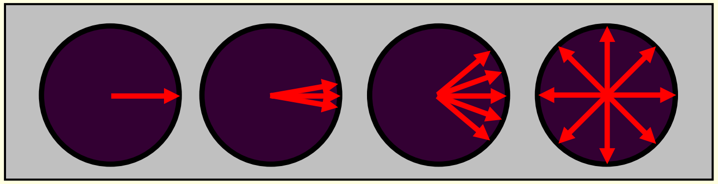
T1 of brain matter



From: Schlegel/Mahr, 3D Radiaton Therapy, 2007

Relaxation time: T2

- Influenced by spin – spin coupling
- Phase coherence in plane:



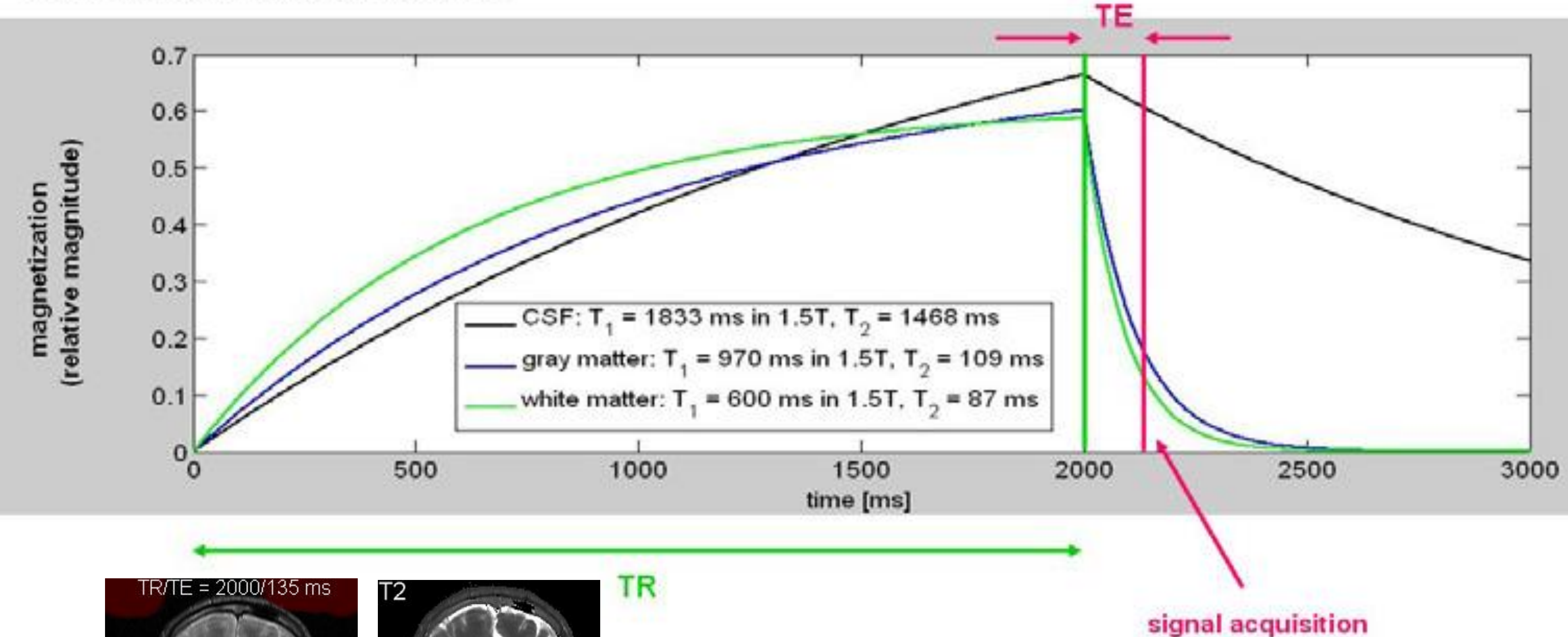
- Sensitive to field and susceptibilty inhomogeneities; without spin echo, signal decays with relaxation time T2*
- Can be measured directly
- Spin echo puls sequence:

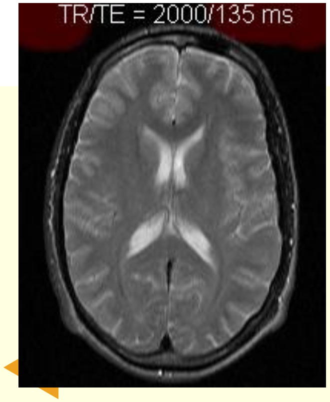
  - long TR (longer than T1), eg. ~2 s
  - TE around T2, eg. ~100 ms

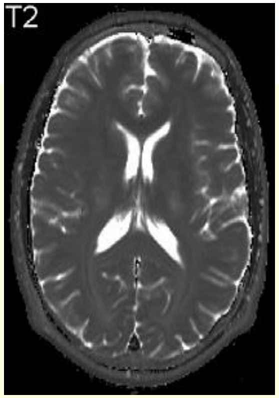
From: Schlegel/Mahr, 3D Radiaton Therapy, 2007

Relaxation time: T2

T2-weighting of brain tissues



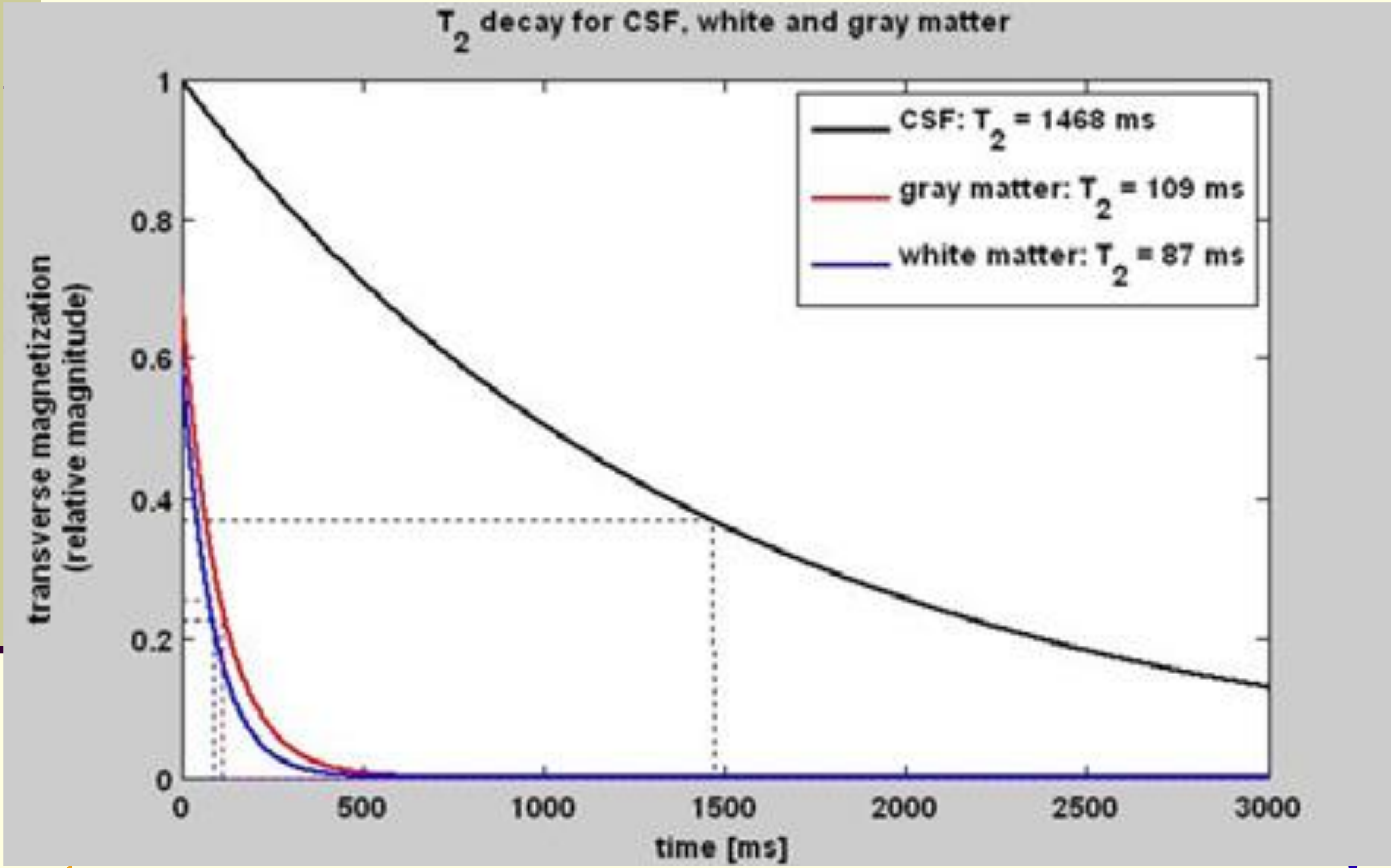




Left: T2-weighted image
Right: image from the actual T2 values of the tissues.

From: Schlegel/Mahr, 3D Radiaton Therapy, 2007

T2 of brain matter



From: Schlegel/Mahr, 3D Radiaton Therapy, 2007

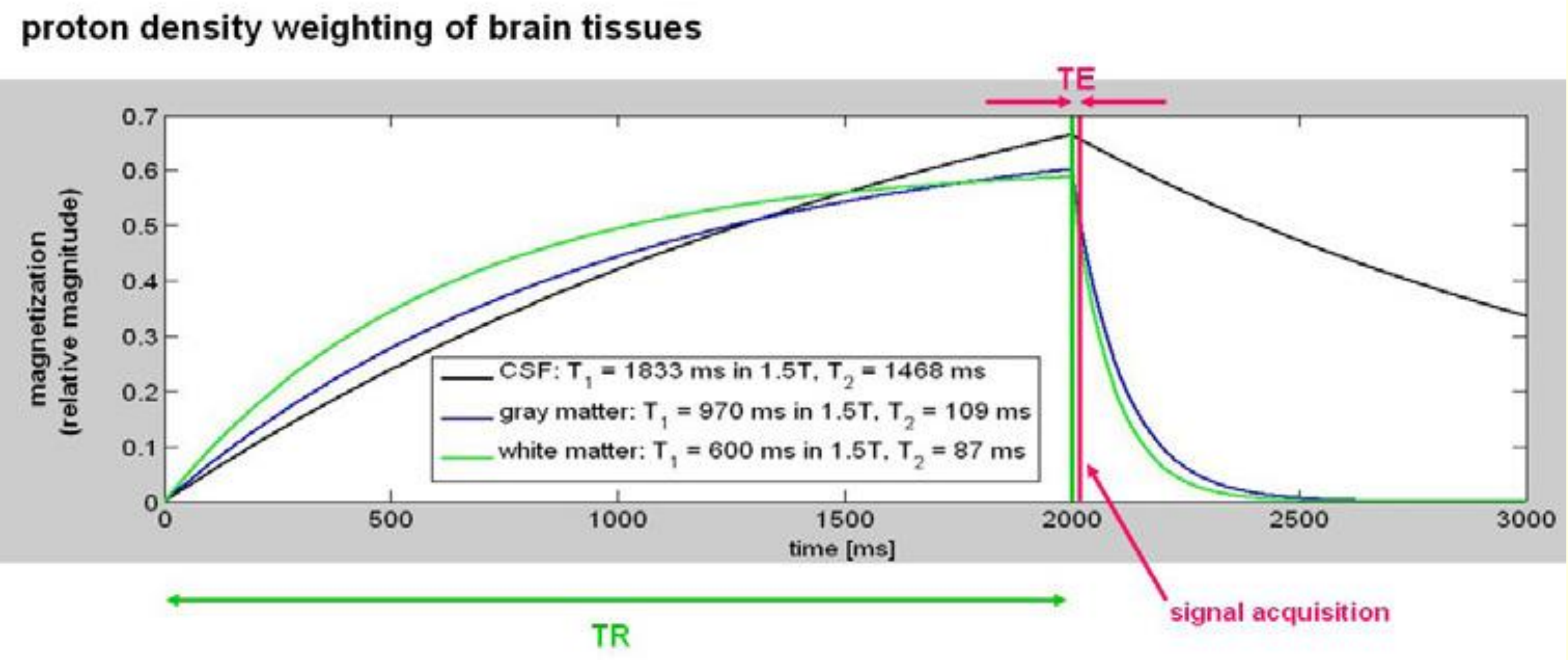
Proton density measurements

- Analysis of proton echo signal amplitude with least influence by T1 or T2
- Spin-echo puls sequence:

  - long TR (longer than T1), eg. ~ 2 s
  - short TE (shorter than T2), eg. ~ 20 ms

- Neither T1 nor T2 have major influence on the signal
  -> signal reflects presence of protons itself!

Proton density measurements



Left: PD-weighted image
Right: image from the actual PD values of the tissues. Cerebrospinal fluid appears dark due to flow artifacts.

From: Schlegel/Mahr, 3D Radiaton Therapy, 2007

Tissue parameters at 1 Tesla: T1, T2 and proton density

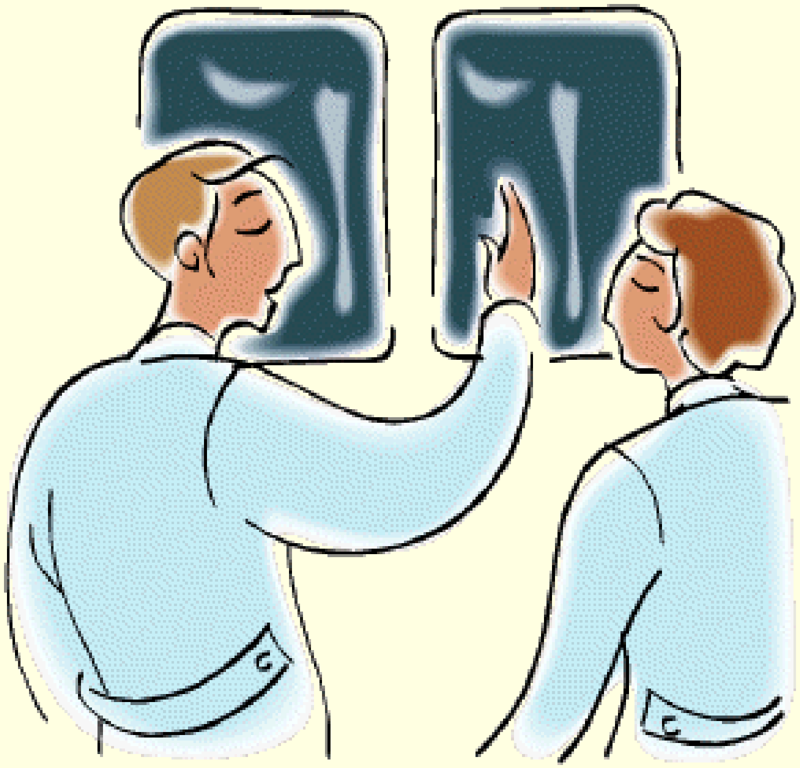
Tissue	T1 / ms	T2 / ms	Proton density
CSF	2400	500	1.0
Gray Matter	920	100	0.85
White Matter	780	90	0.70
Adipose Head	260	85	1.23
Adipose Abdomen	260	85	1.54
Spleen	780	62	1.15
Kidney	650	58	0.99
Liver	490	43	1.0
Tumor	500	100	1.0

From: http://www.lems.brown.edu/~msj/cs292/assign2/mr.html

MR puls sequences

- **Spin echo (SE)**: 180° pulse refocusses spins, less influenced by field inhomogeneities, long measuring times (minutes for a slice)
- **Echo planar imaging (EPI)**, uses no 180° puls; very fast: image slice in 0.1 s; sensitive to field inhomogeneities due to $T2^{*}$; accumulated phase shift results in pixel shift for fat; used for diffusion, perfusion and functional (BOLD) imaging
- **Inversion recovery (IR)**: Heavy T1 weighting; uses 180° puls to invert magnetization, 90° puls after inversion time TI to measure. Can be used to cancel fat signal (chemical shift at 1.5 T: 220 Hz)
- **Gradient echo (GE)**: no 180° puls to refocus spins, only gradient fields for read out
- **FLASH** (Flip Low Angle SHot), flip longitudinal magnetization only typ. 30°, TR can be much shorter!

Clinical applications of MR



Clinical application:
Soft tissue / brain

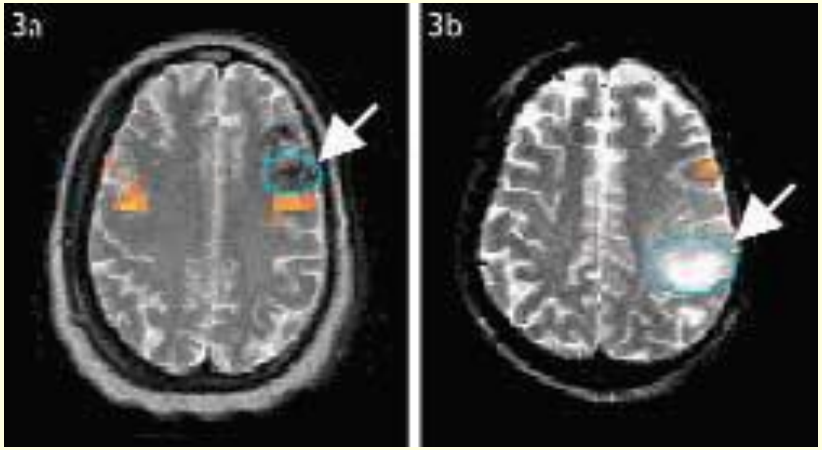
7 Tesla MR,

32-channel head array Rx coil and detunable birdcage Tx coil

2D FLASH,
TR/TE= 500/30
0.22 x 0.22 x 1mm^3 (48nl)
8 min TA

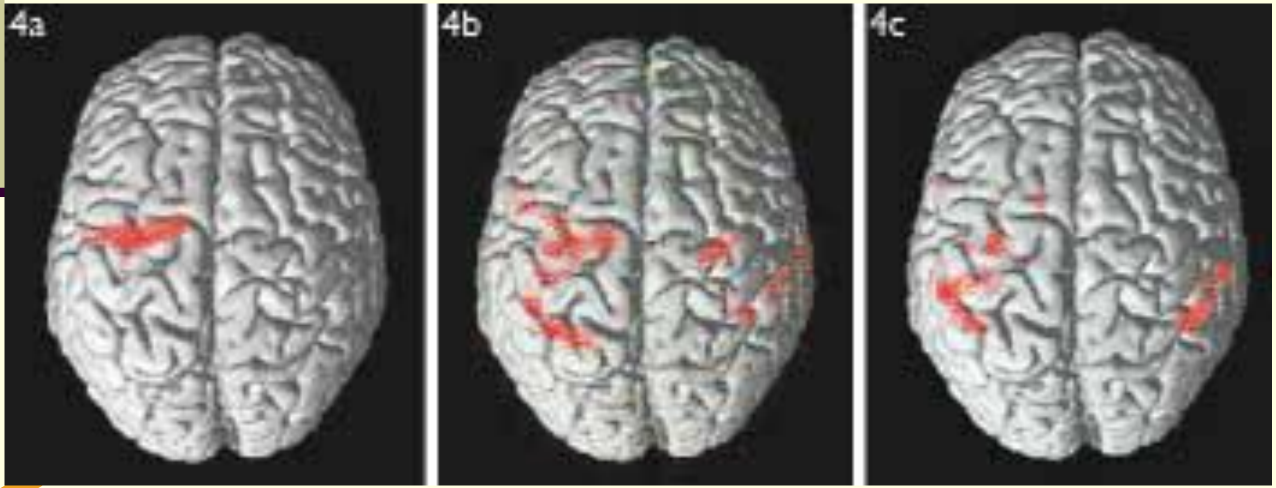
Courtesy MGH, Boston, USA

Clinical application: (functional) fMR – BOLD scans



fMRI in preoperative planning

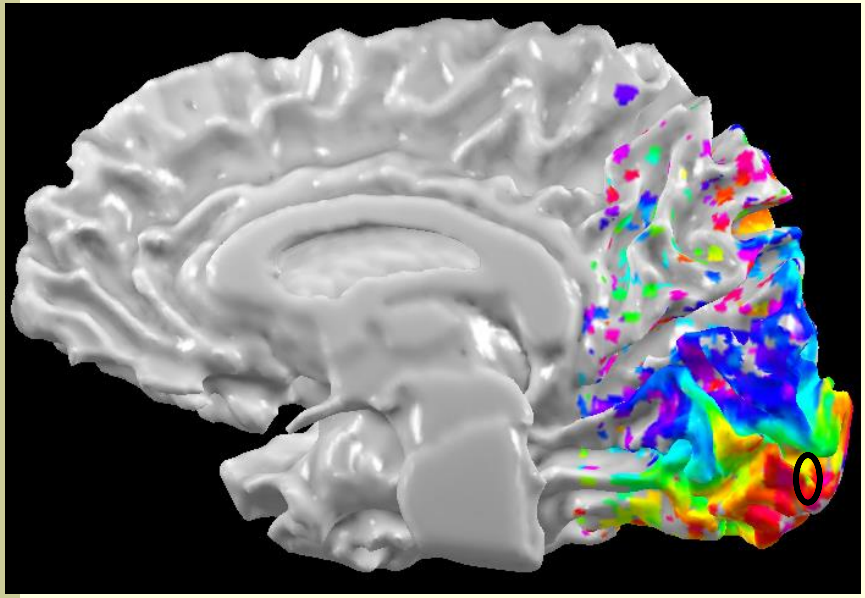
Left: Preoperative planning for biopsy of a lesion (arrow)). Significant bilateral language activation
Right: fMRI determined the language area and relation to tumor for patient with glioblastoma (arrow).



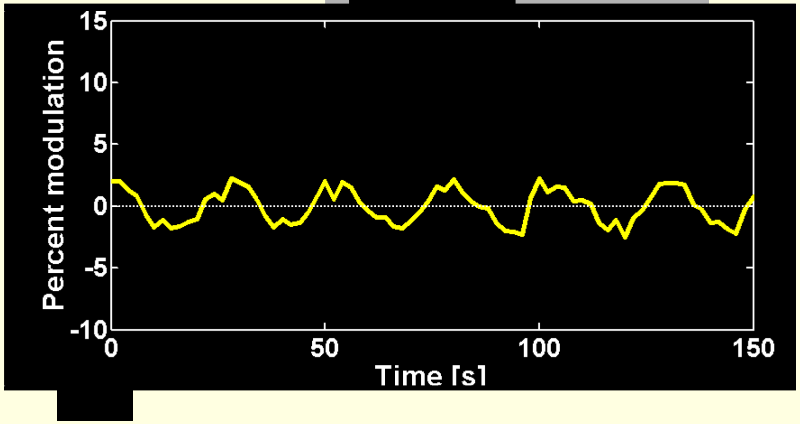
Functional reorganization of motor area

Patients with short term and long term spinal cord injury (SCI).
Left: Control.
Middle: Two years after SCI.
Right: > 10 y after SCI.

Clinical application: (functional) fMR – BOLD scans

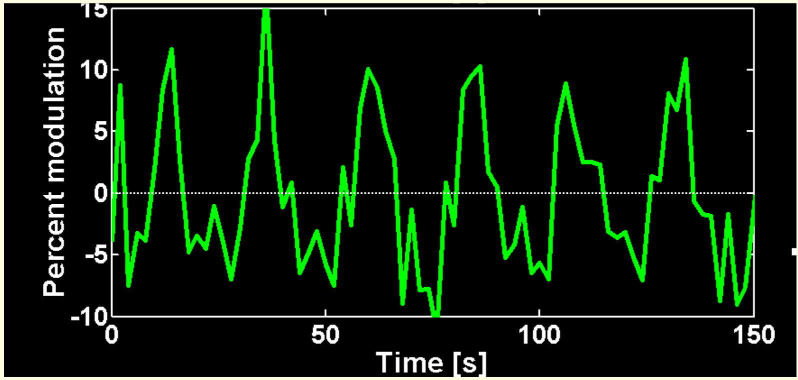


3T



Voxel volume: 13.2mm³

7T



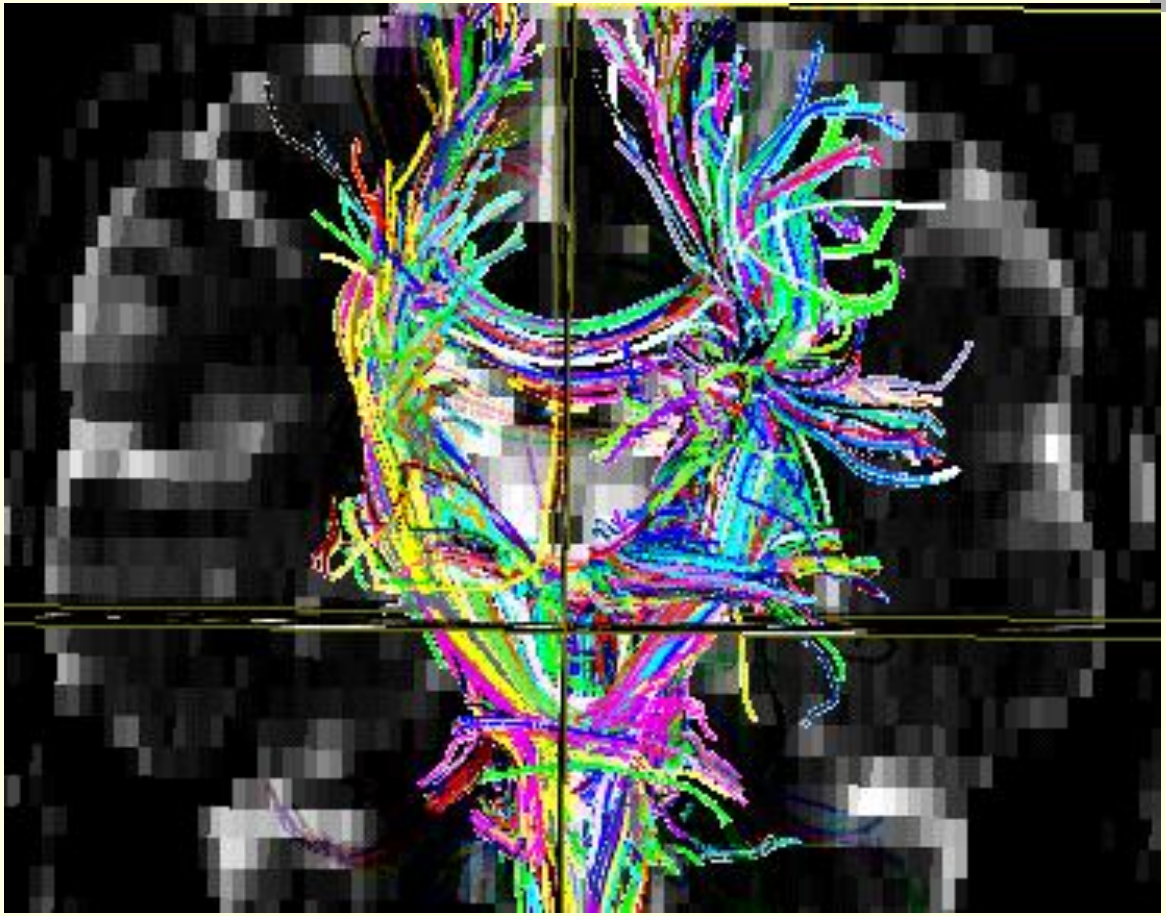
Voxel volume: 23.7mm³

Time series from activation close to the occipital pole

(7T and 3T data scanned under optimal conditions)

 Courtesy Andre Brechmann, IFN, Magdeburg

Clinical application: DTI fiber tracking



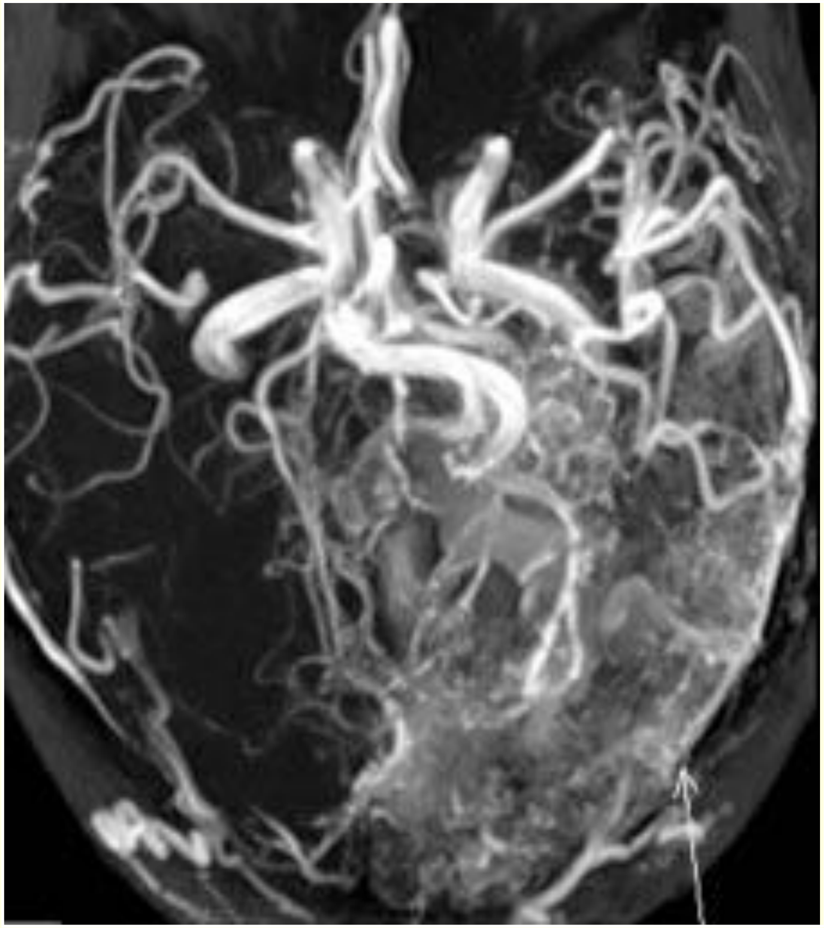
Diffusion Tensor Imaging

Courtesy W. Grodd, Tübingen, Germany

Clinical application: MR phase contrast angiography

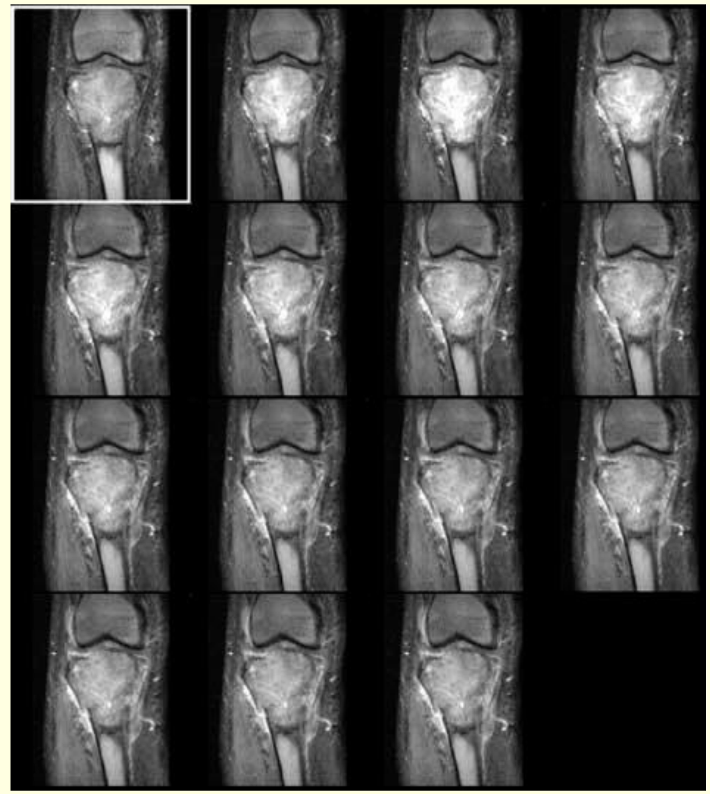
Problems:

- very long measuring times
- sensitive to patient motion
- max. flow velocity must be known (VENC)
- Pulsation artefacts (arterial MRA without ECG-trigger)
- long echo times cause signal extinction for turbulent flow



Courtesy F. Schick, Tübingen

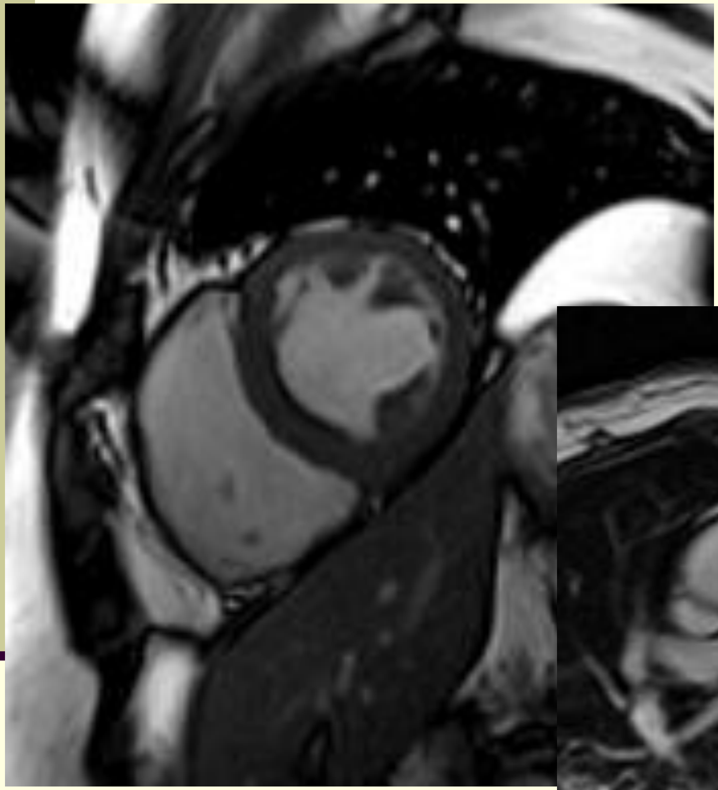
Contrast enhanced MRA

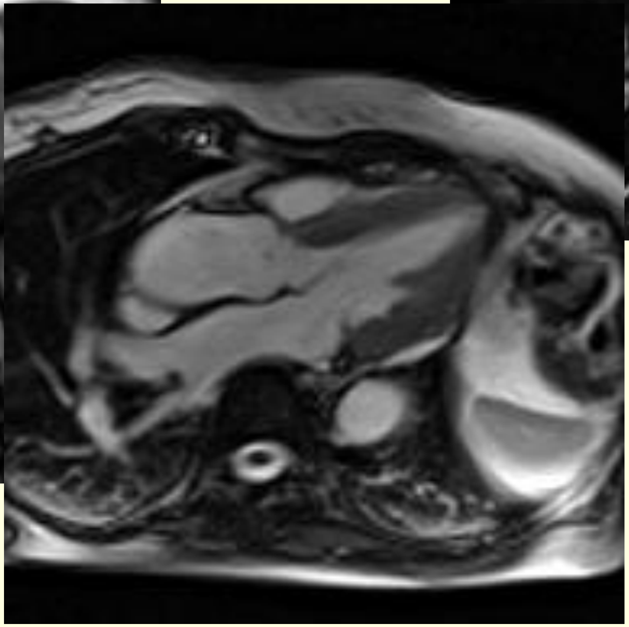


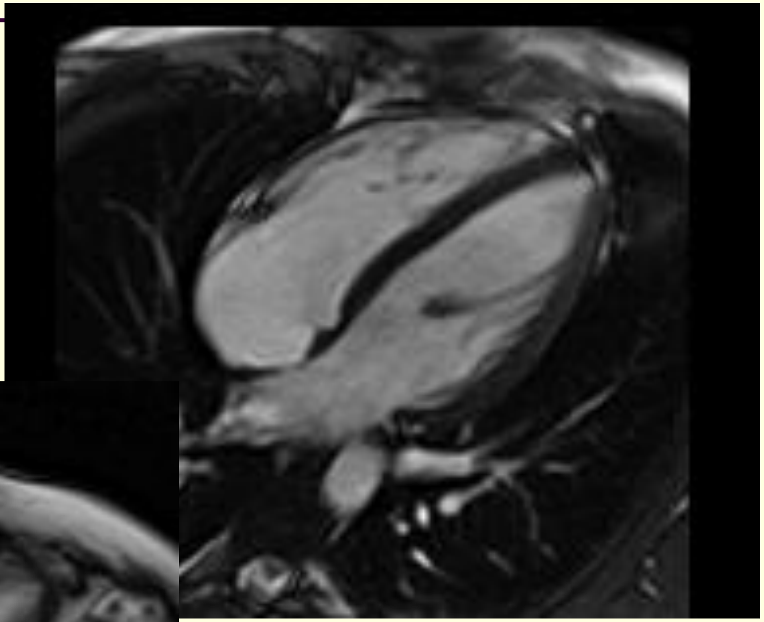
Use of the T1-shortening caused by contrast agent

Courtesy F. Schick, Tübingen

Clinical application: Cardiac imaging



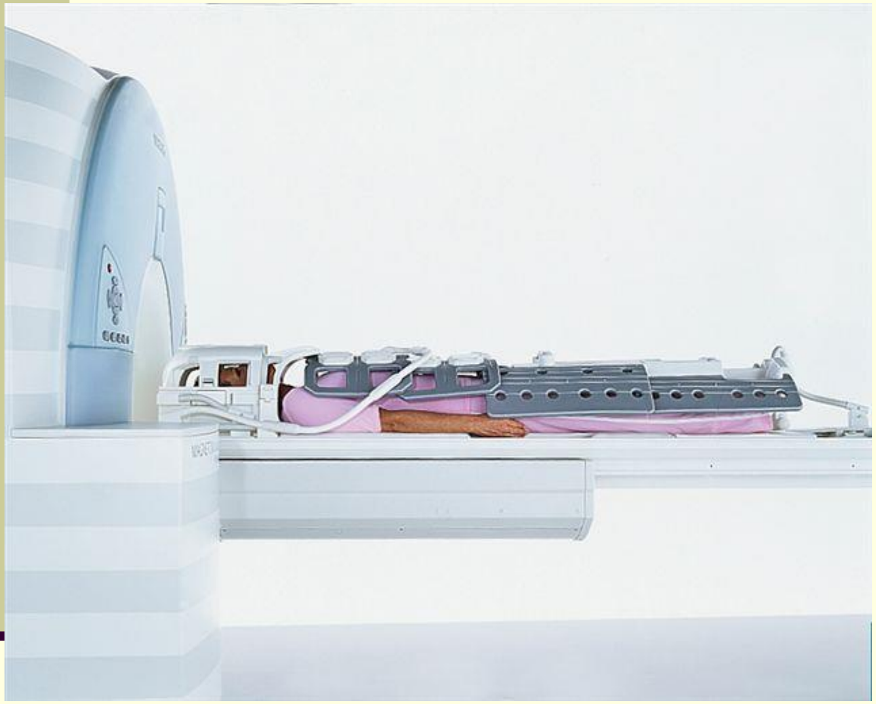


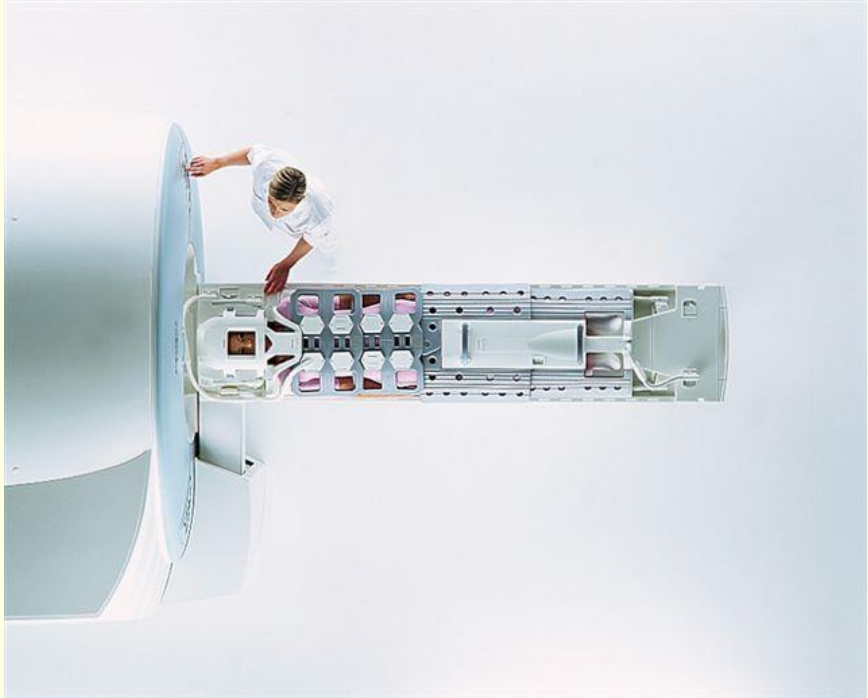


ECG triggered resampling

Courtesy Siemens

Whole body-MRI





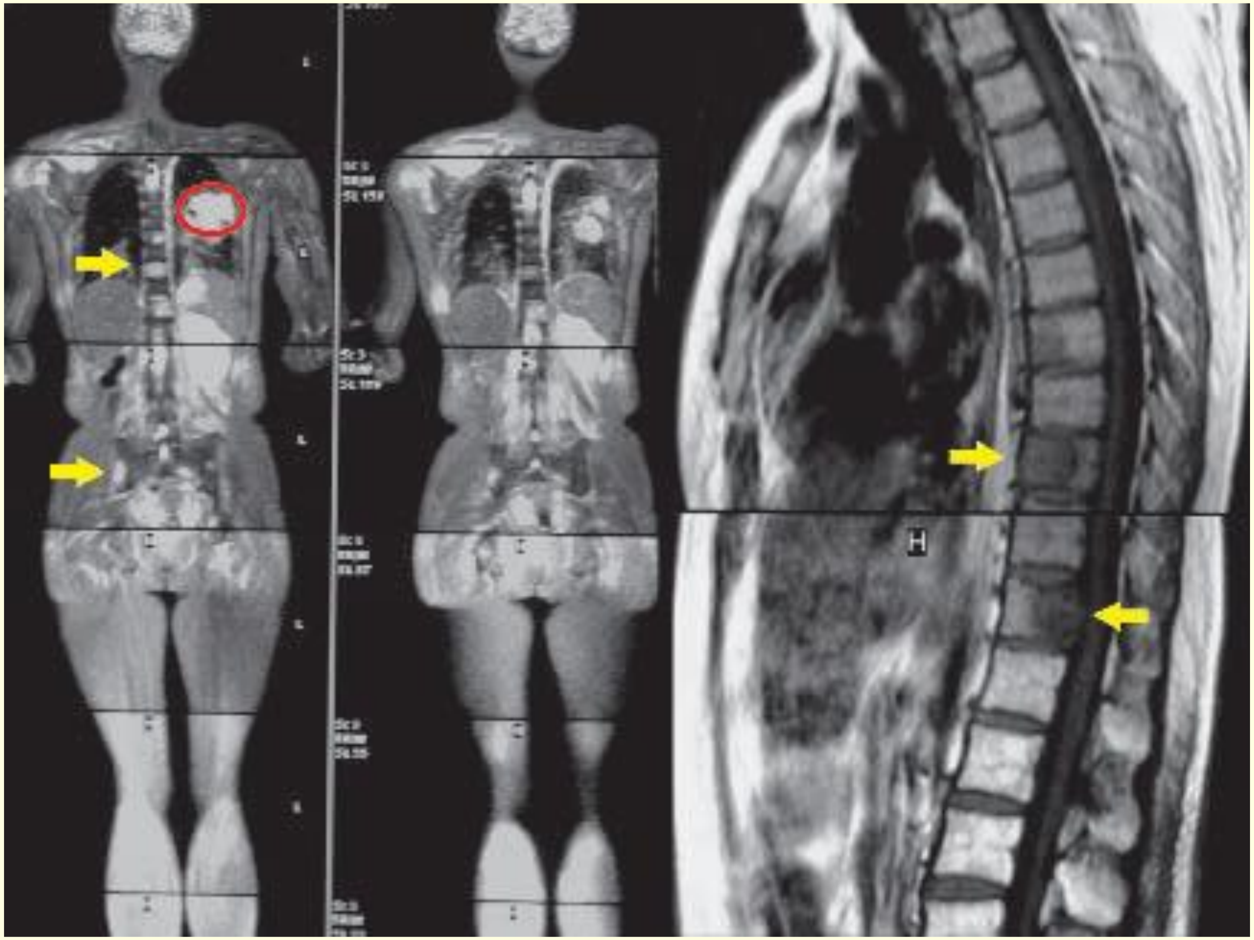
Siemens TIM: movement of couch in sections of < 40 cm

(**T**otal **I**maging **M**atrix, max. 2.05 m)

<u>Experimental:</u> continous movement of couch during scanning

Courtesy Siemens

Clinical application:
Whole body tumor screening



TurboSTIR whole-body MRI

Left + middle: metastatic breast Carcinoma, metastasis in the left lung with diffuse metastatic deposits in spine and pelvis.

Right: Sagittal T1-weighted image confirms presence of spinal metastases.

From: MedicaMundi,47/3 Nov 2003

Whole body-MRI

Morphology

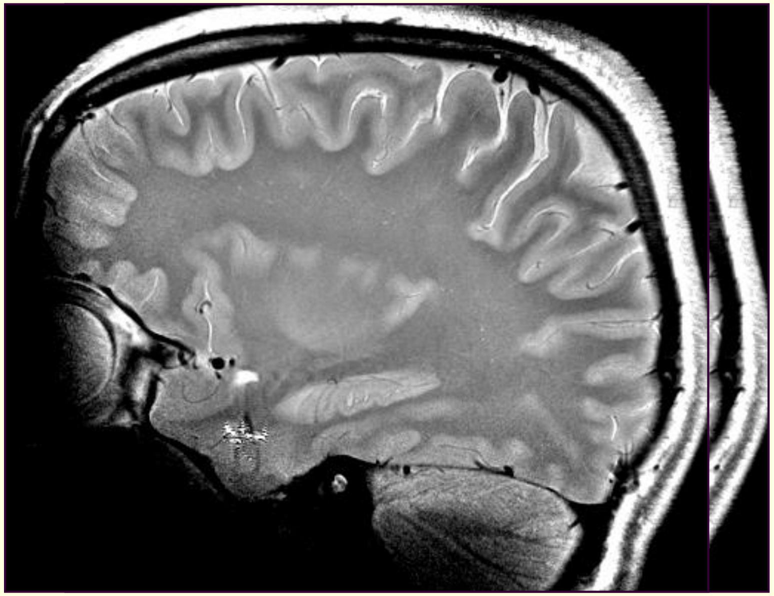
+

Cardio-vascular system

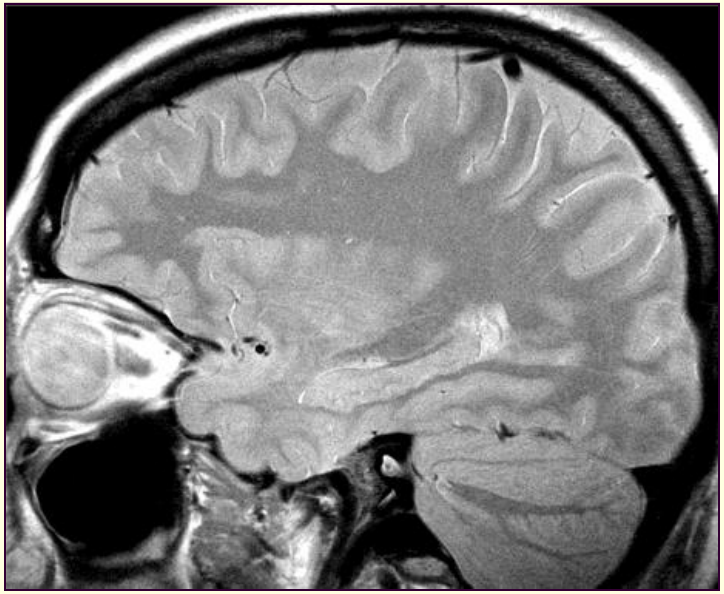
Courtesy F. Schick, Tübingen

High field MR: Signal to noise ratio

7T



3T



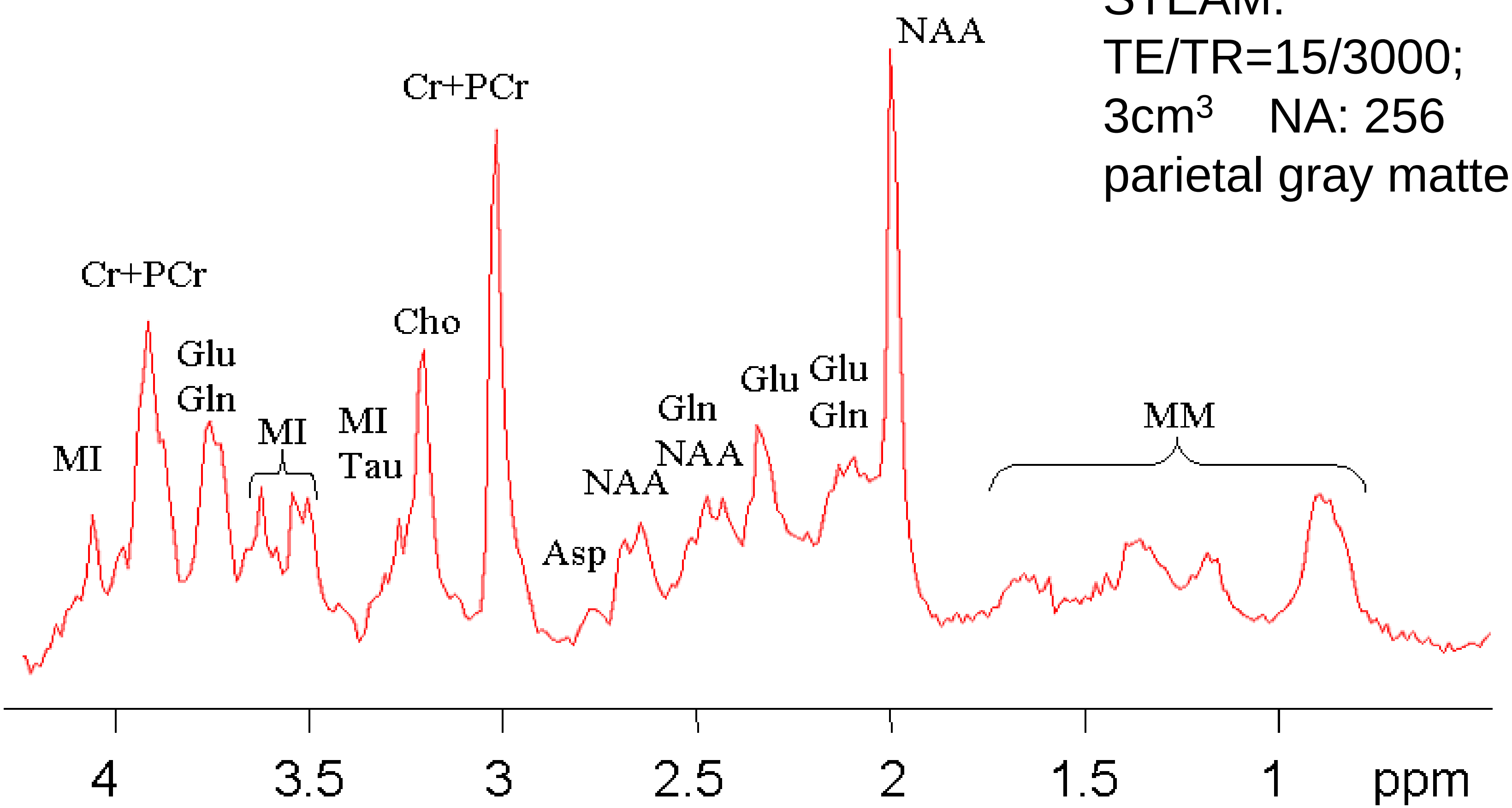
TSE, 11 echoes, 7 min exam, 20cm FOV, 512x512 (0.4mm x 0.4mm), 3mm SL

7T	**3T**
white matter SNR = 65	white matter SNR = 26
gray matter SNR = 76	gray matter SNR = 34

Courtesy of LL. Wald et al., MGH NMR-Center, Boston, USA

Clinical application high field MR: Spectroscopy

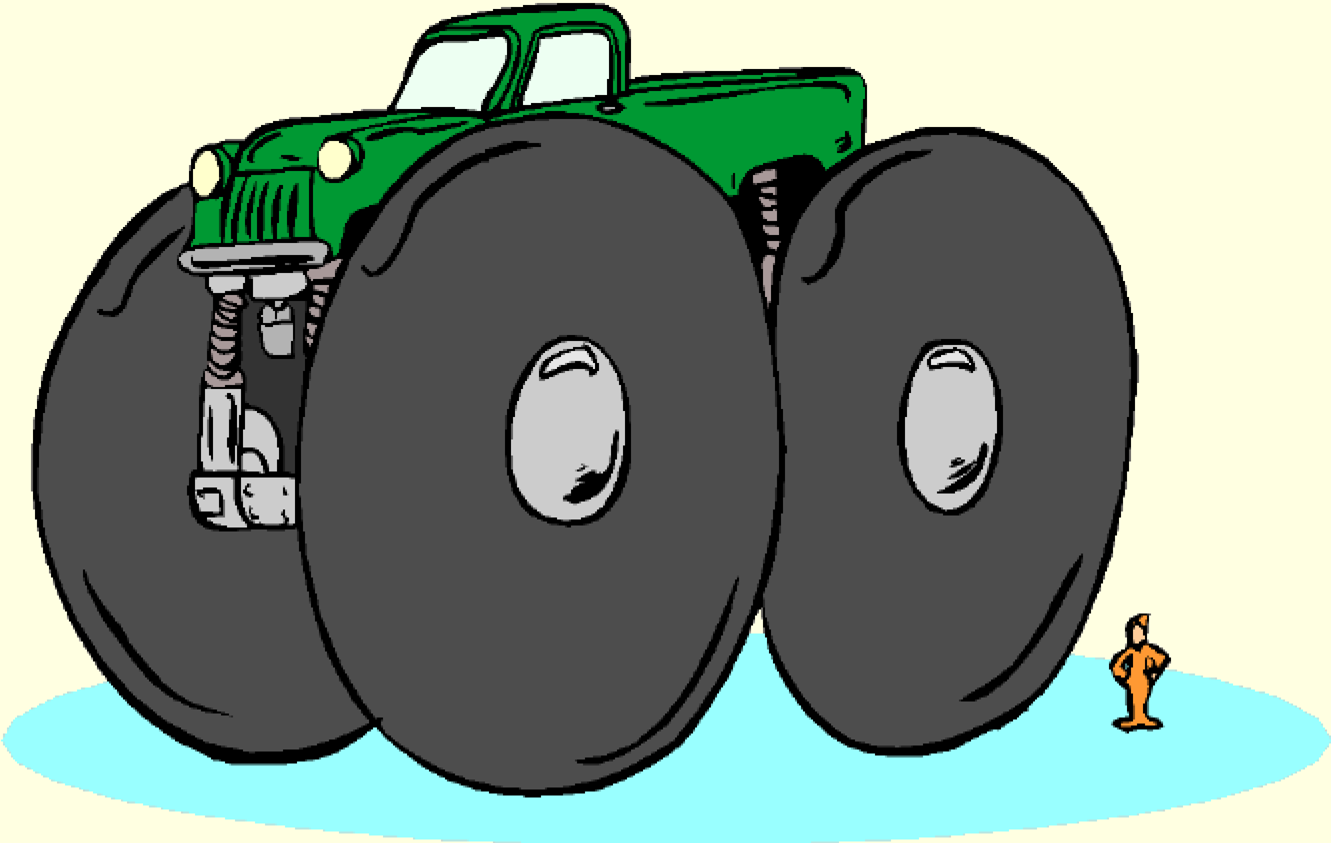
STEAM:
TE/TR=15/3000;
3cm^3 NA: 256
parietal gray matter



Single Voxel Spectroscopy, rhesus macaque

Courtesy, CMRR, Minnesota, USA

Magnets: stronger, wider, bigger



Future of MRI: 9.4 T installed 2007 at MPI Biological Cybernetics,Tübingen

Magnet:
Weight (without He): 47 t;
Size: 3.67 m long,
4.56 m heigh,
2.86 m wide.

Usabel bore diameter: 30 cm

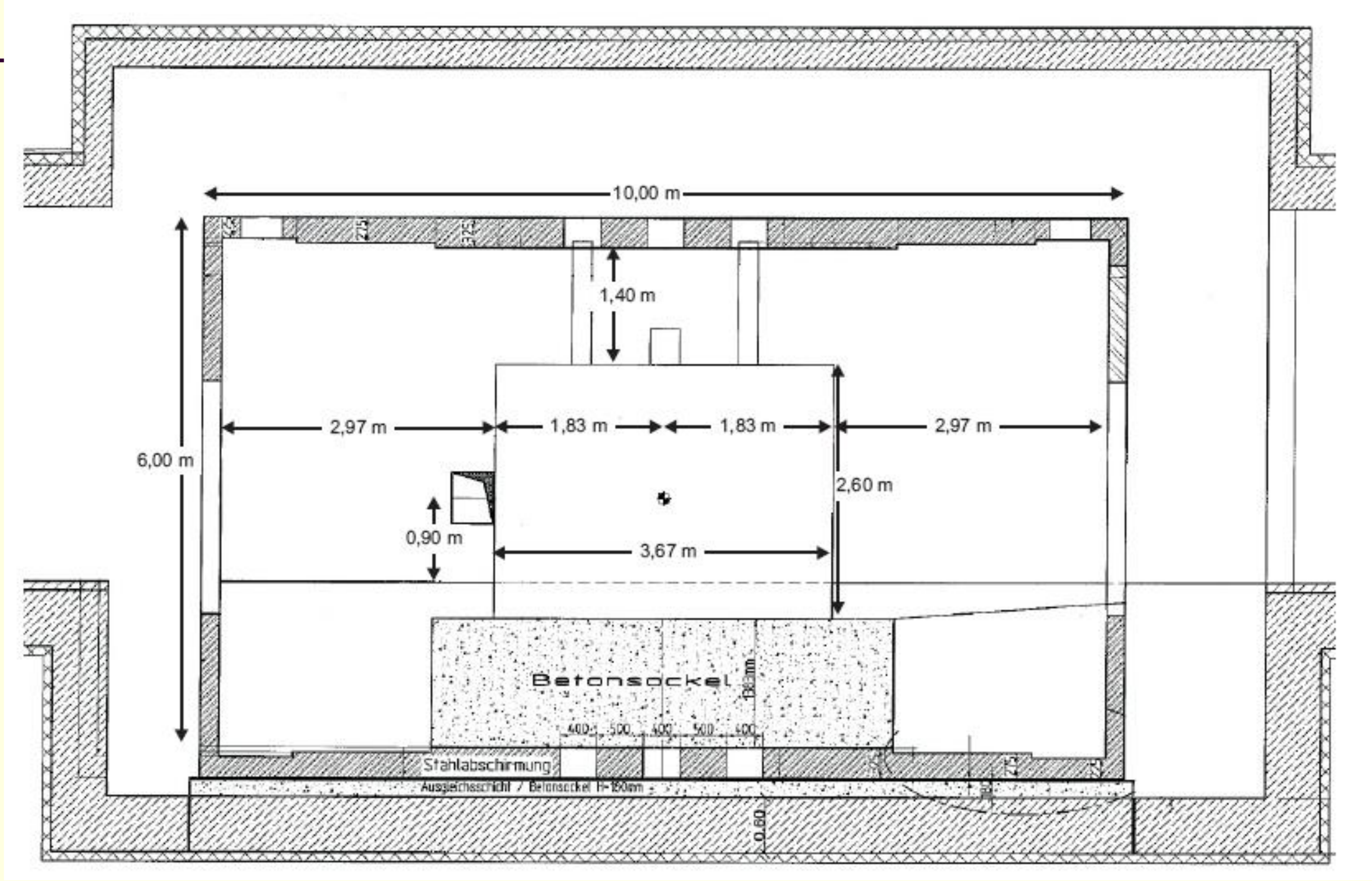
Steel shielding cage: 602 t
(10 x 6 x 6 m^3)

Field homogeneitiy:
better 0,03 ppm
(0,3 μT),
Shim coils: 14 superconducting
8 resistive

Energy stored in magn. Field at 9.4 T: ~78 MJ

Courtesy R. Pohmann, Tübingen

9.4 Tesla MRI: magnetic shielding



Shielding cage of 602 t steel, wall thickness up to 33 cm

Eberhard-Karls-Universität
UKT
Universitätsklinikum Tübingen

Courtesy R. Pohmann, Tübingen

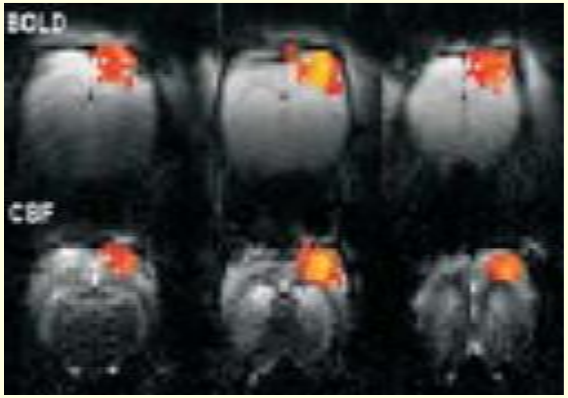
medphys

16.4 Tesla small animal MRI

Magnet:
Weight (without He): 22 t;
Size: 3.47 m long, 2.88 m heigh and 2.10 m wide,
steel shielding cage: 220 t
(8 x 4.50 x 4.50 m^3)

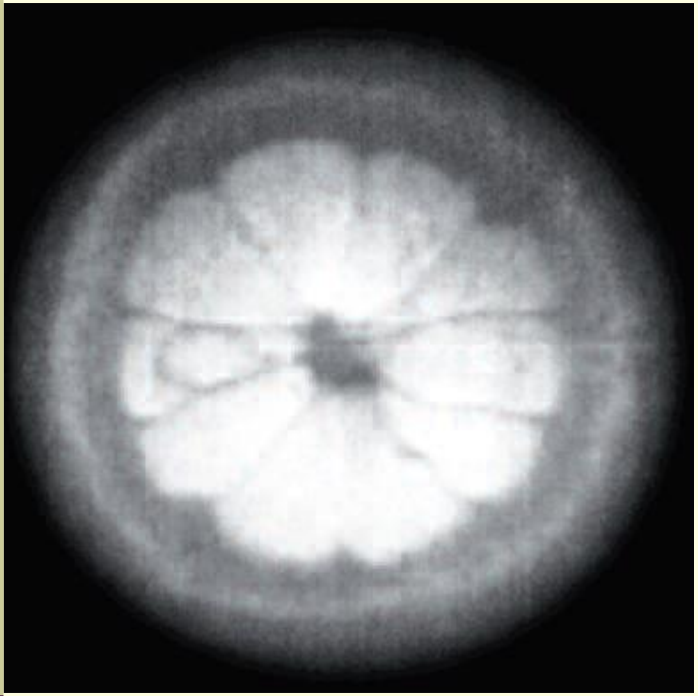
Energy stored in magn. Field eq. LearJet at 150 km/h!

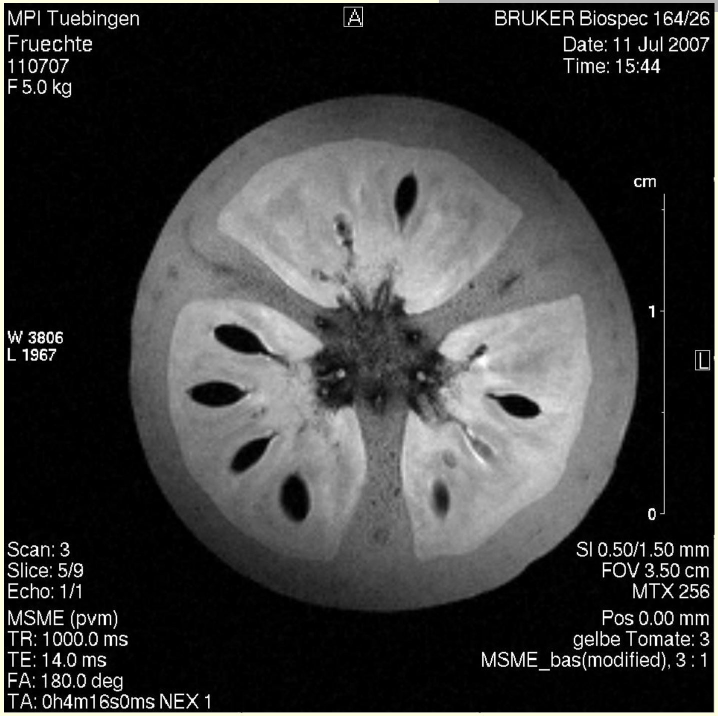
fMRI signal bei 9,4 T of brain of a rat wiht electric stimulation of front leg.
Silva et al. J Cereb Blood Flow Metab 1999;19:871–879.



Courtesy R. Pohmann, Tübingen

High resolution MR images





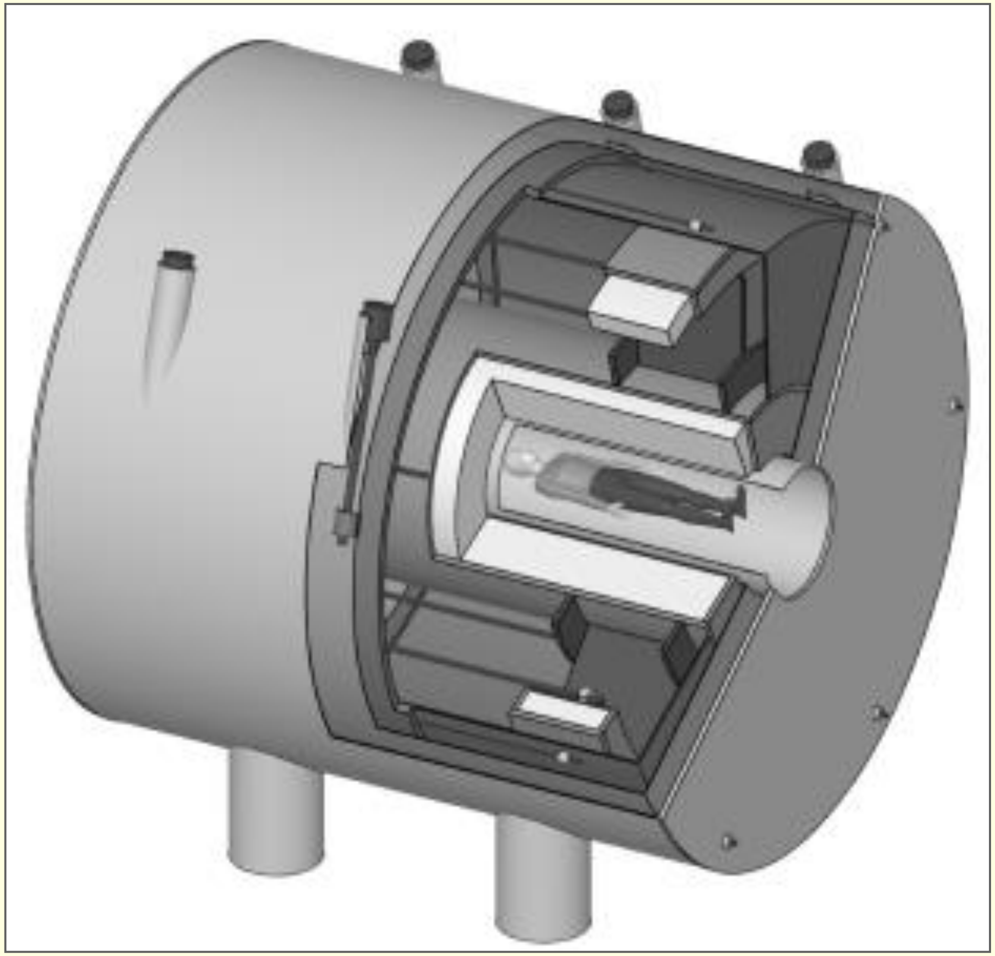
Historic MR image:
The Nottingham Orange
(1978)

The Tübingen Tomato
(16.4 Tesla, resolution ~100µm)

From: MedicaMundi 50/1 2006/5

Courtesy R. Pohmann, Tübingen

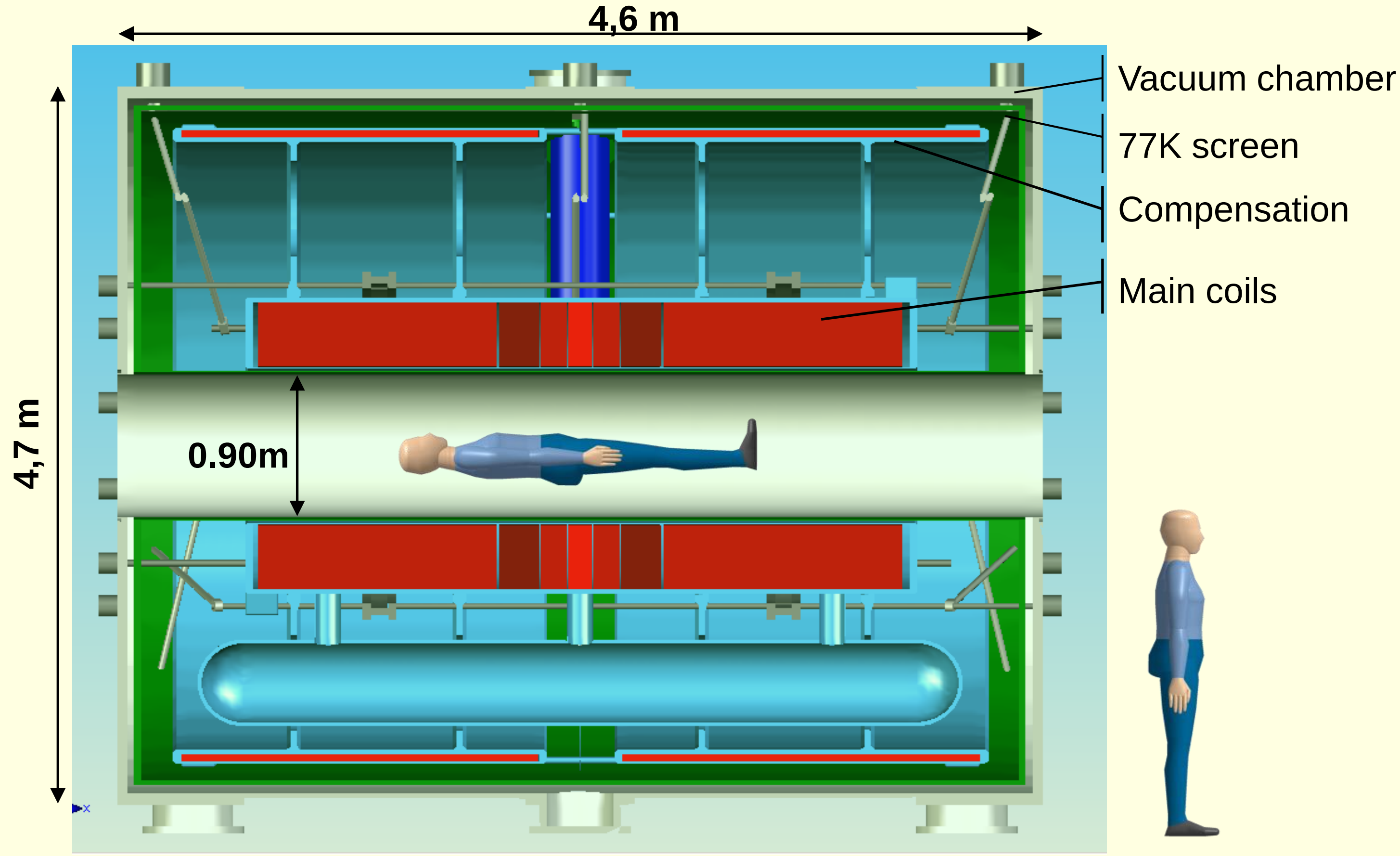
The NeuroSpin project at CEA, France:



wide bore 11.7 T magnet

Courtesy CEA, Paris, France

The NeuroSpin project: wide bore 11.7 T magnet

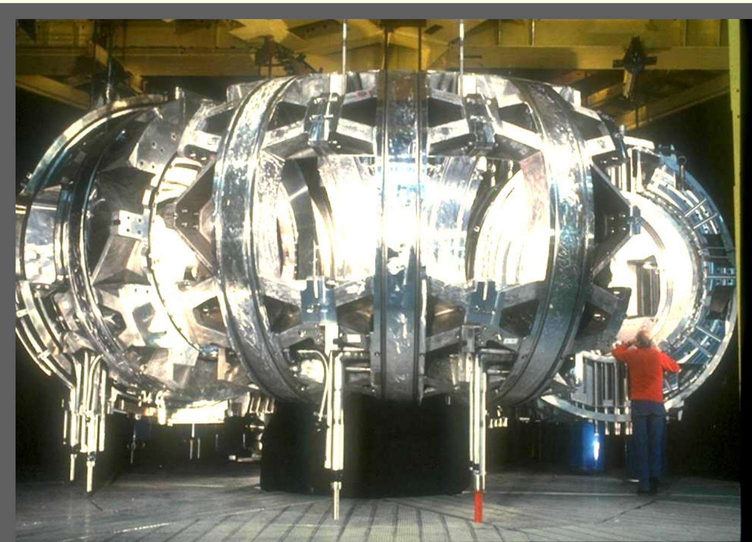


Courtesy CEA, Paris, France

Superconducting Supermagnets „Monster Magnets“

Magnet installations for physics research by Alstom

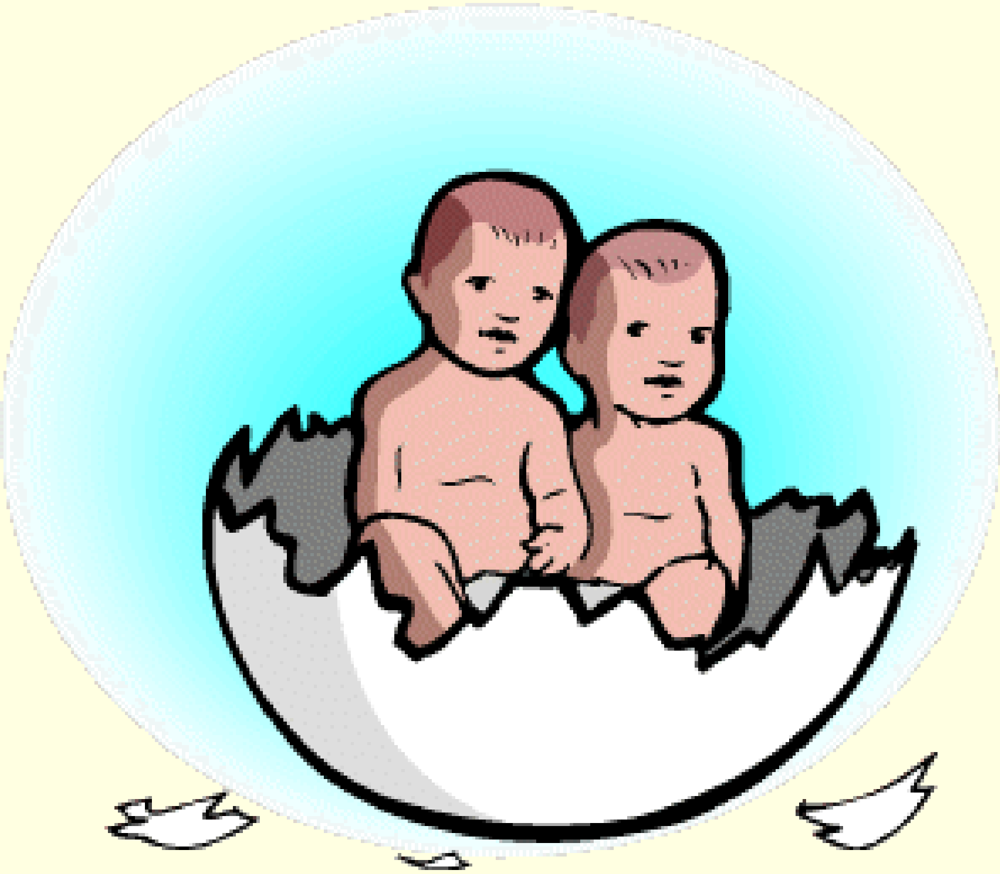
12m long magnet
(6m bore diameter)



600 MJ « Tore » magnet
(nuclear fusion)

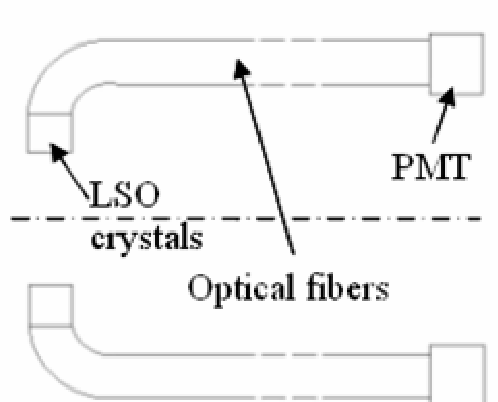
Courtesy CEA, Paris, France

The new combination: PET-MR

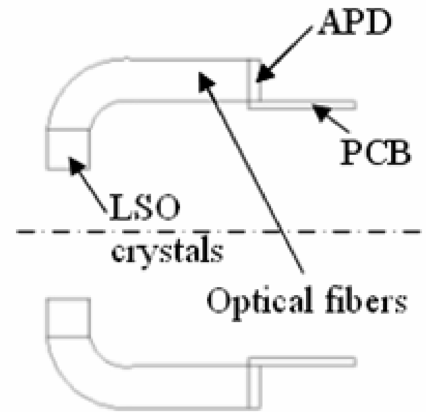


Approaches for Combined PET/MR

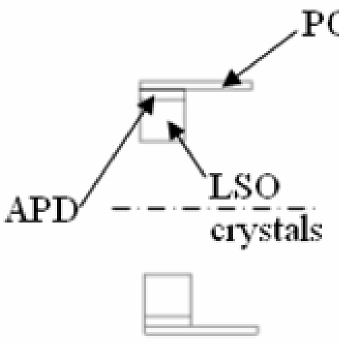
Fibers coupled to PMTs	APDs coupled via short fibers	APDs directly coupled



LSO: LuthetiumOrthoSilicate
PMT: Photomultiplier



APD: AvalanchePhotoDiode



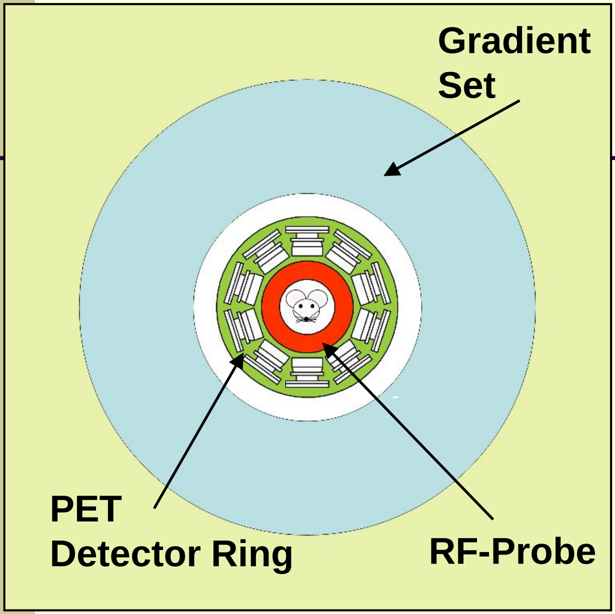
Fibers coupled to PMTs
- Least interference with MRI
- Limited axial FOV

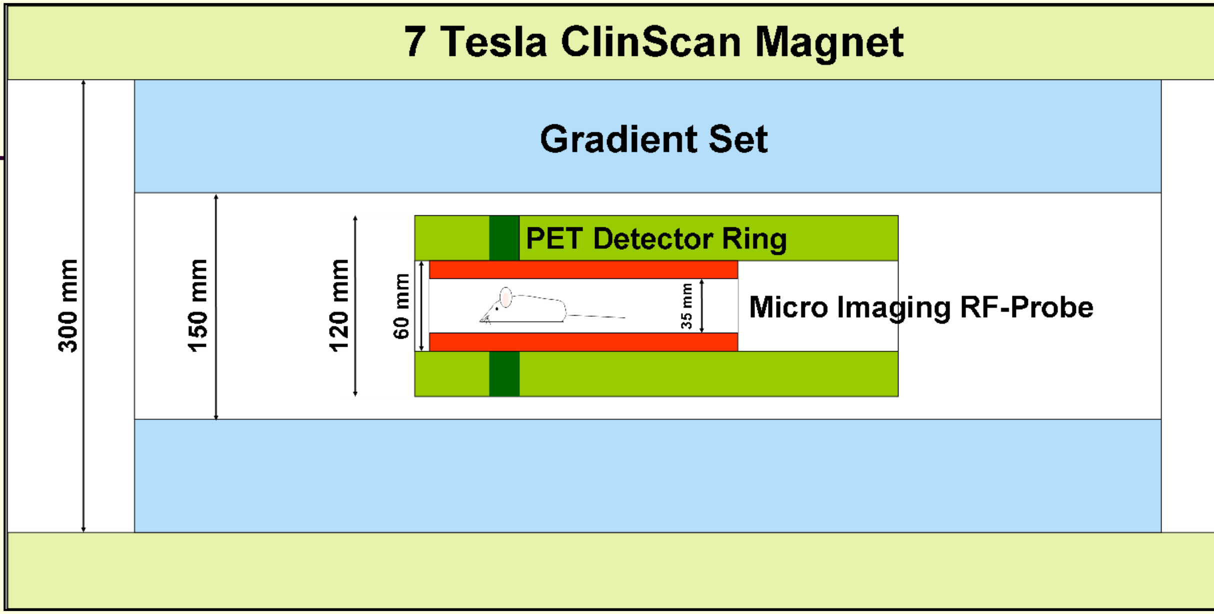
APDs coupled via short fibers
- No metals in MRI FOV
- Limited axial FOV

APDs directly coupled
- Compact design
- Metals in the field

Courtesy B. Pichler, Tübingen

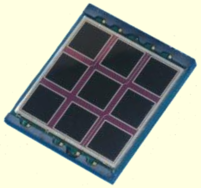
Design of the PET Insert

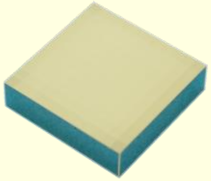




<u>Configuration of the PET Insert:</u>

- 12 x 12 LSO crystal blocks (1.6 x 1.6 x 4.5 mm^3)
- 3 x 3 monolithic APD arrays (Hamamatsu).
- 10 detector blocks
- 19 mm axial FOV, 38 mm transaxial FOV
- micro Imaging RF-probe (35mm) (Bruker BioSpec) is used for MR imaging

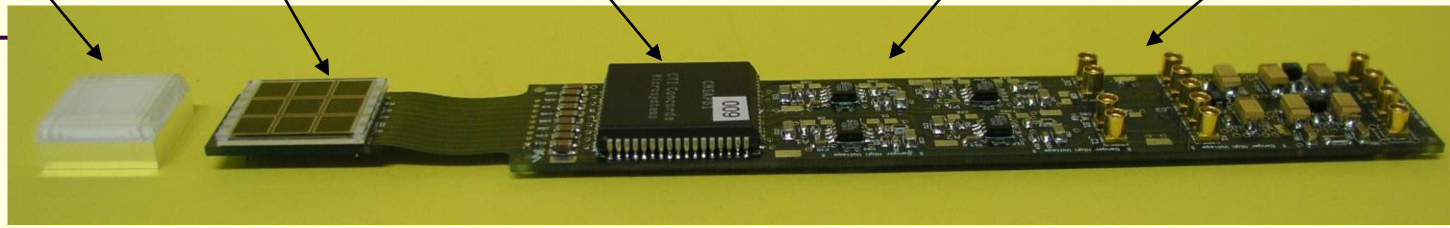




Courtesy B. Pichler, Tübingen

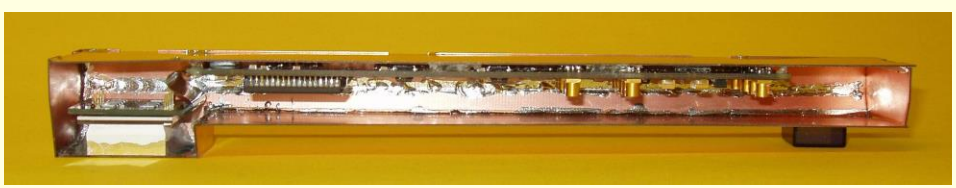
Detector Module Design

LSO block 3x3 APD array preamplifier buffer connectors



- 6-layer printed circuit board with flexible tip
- non magnetic parts used for population
- integrated charge sensitive preamplifier (Siemens) performance
  - rise time of 13.4 ns (test pulse with 12 ns rise time)
  - gain of 0.93 µV/e-
  - rms noise 598 + 9.2 e- /pF (150 ns shaper)

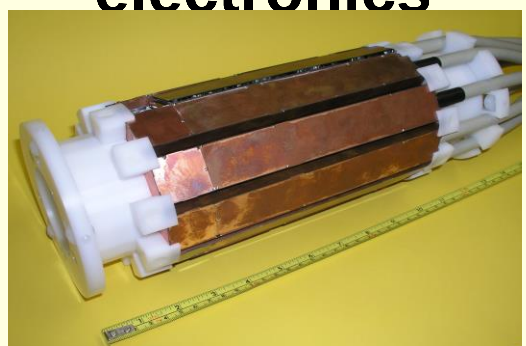
- 10 µm copper covered PCB material used as shielding.



Courtesy B. Pichler, Tübingen

Full PET-MR System

- 7 Tesla ClinScan MR system (Bruker, Germany)
- 10 detector blocks (at room temp.)
- Readout with Siemens-Quicksilver electronics



PET scanner parameters

Crystal size	1.6 x 1.6 x 4.5 mm³
Crystal material	LSO
Number of detector blocks	10
Number of crystals per block / ring / total	144 / 120 / 1440
FOV axial / transaxial	19 mm / 38 mm
Coincidence timing resolution	8 ns
Resolution	< 2 mm

Courtesy B. Pichler, Tübingen

System Performance Test

Influence of the PET insert on the MR Imaging

TSE Sequence
TR 2500 ms
TE 43 ms
256 x 256

$$\text{Homogeneity} = 100x\left(1 - \frac{R_{Max} - R_{Min}}{R_{Max} + R_{Min}}\right)$$

Phantom images were acquired with and without PET insert

Sequence	TR	TE	SNR			Homogeneity		
	ms		PET	no PET	change	PET	no PET	change
TSE	2500	43	50	61.8	-19%	85%	87%	-3%
GE	300	7.1	19.3	24.7	-22%	79%	77%	3%
GE	100	10	18.1	17.3	5%	70%	65%	8%

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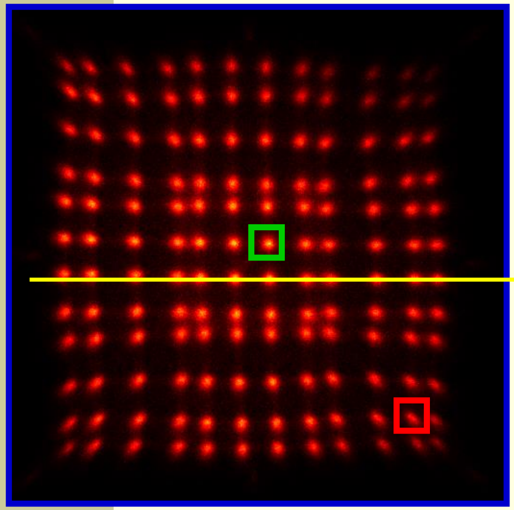
Courtesy B. Pichler, Tübingen

Phys

System Performance Test

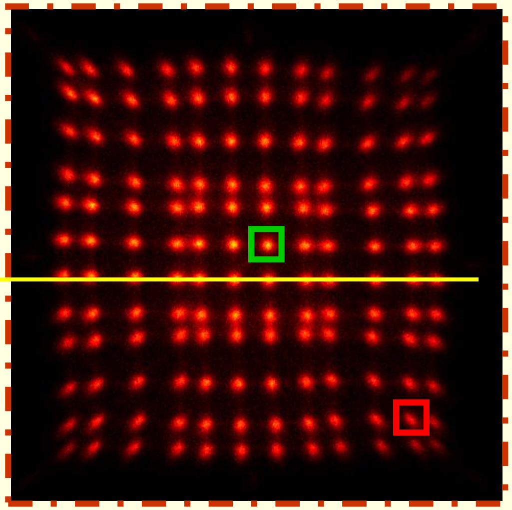
Influence of the MR on the PET insert

Outside MR

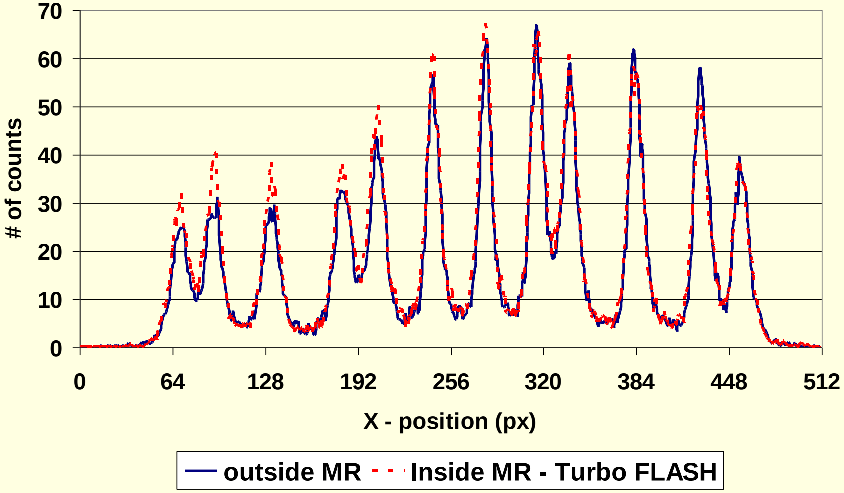


17.8% / 19.3 %

Inside MR



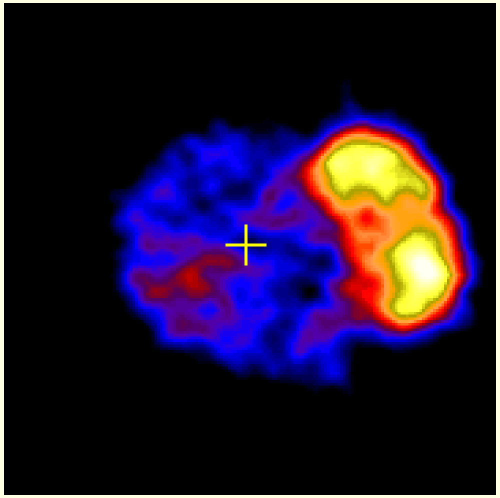
18.0 % / 20.1 % Energy resolution (@511 keV)



Position profiles of the PET insert were acquired outside and during MR imaging.

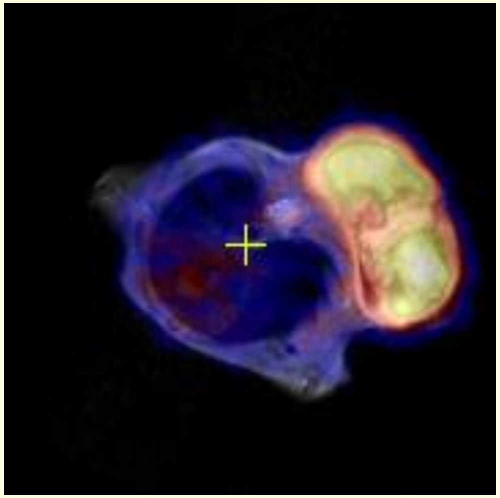
➢ **No noticeable influence could be seen from the flood images**

In-vivo static PET-MR Imaging



PET

MR



PET/MR

- 17 week old female Balb/C mouse – bearing CT26 tumor
- 280 µCi [^{18}F]-FLT (Fluor-Thymidine) Tracer
- 30 min scan after 60 min uptake (190 uCi)
- TSE MR sequence TR = 2770; TE = 42; 4 aver.;
  256 x 256 matrix; 1mm slice thickness

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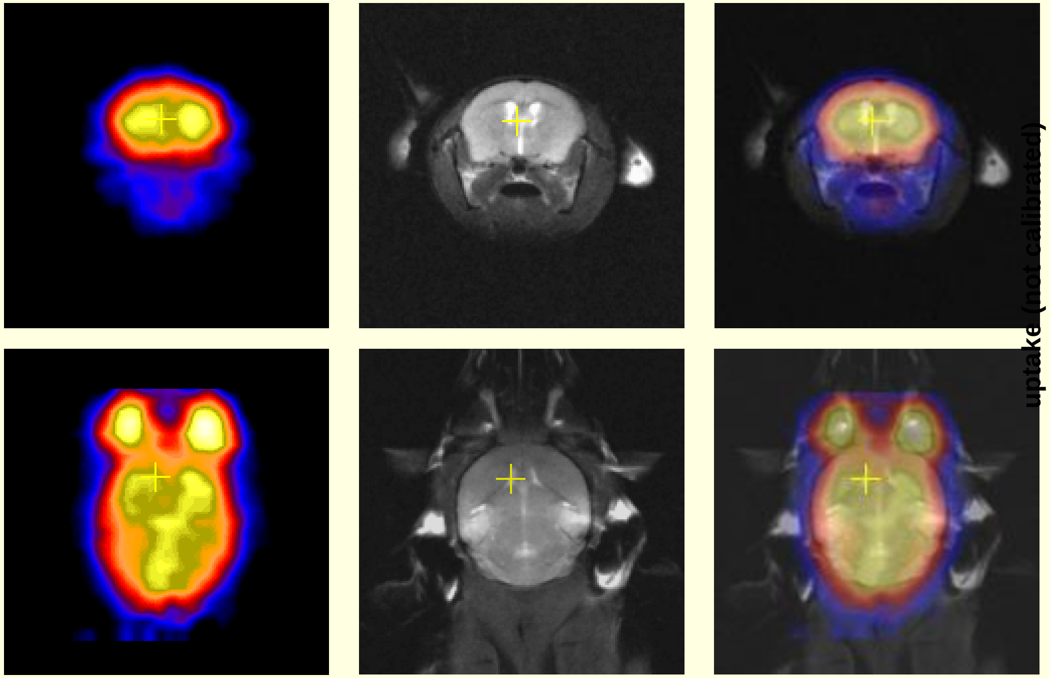
Courtesy B. Pichler, Tübingen

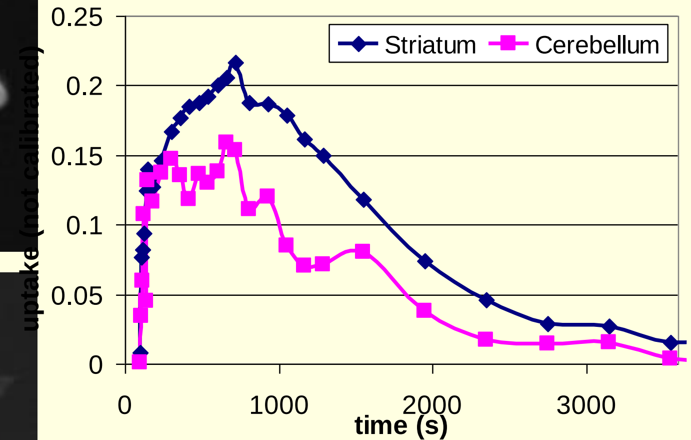
In-vivo Dynamic PET-MR Imaging

transvers

coronar



PET MR PET/MR



Logan plot of TACs shows BP of 0.8

- **female C57BL/6 mouse**
- **580 µCi [^{11}C]-Methylphenidate** **(Dopamine transporter ligand)**
- **60 min dynamic acquisition**
- **TSE MR sequences** **TR = 2770; TE = 42; 256 x 256 matrix; 1mm slice**

Courtesy B. Pichler, Tübingen

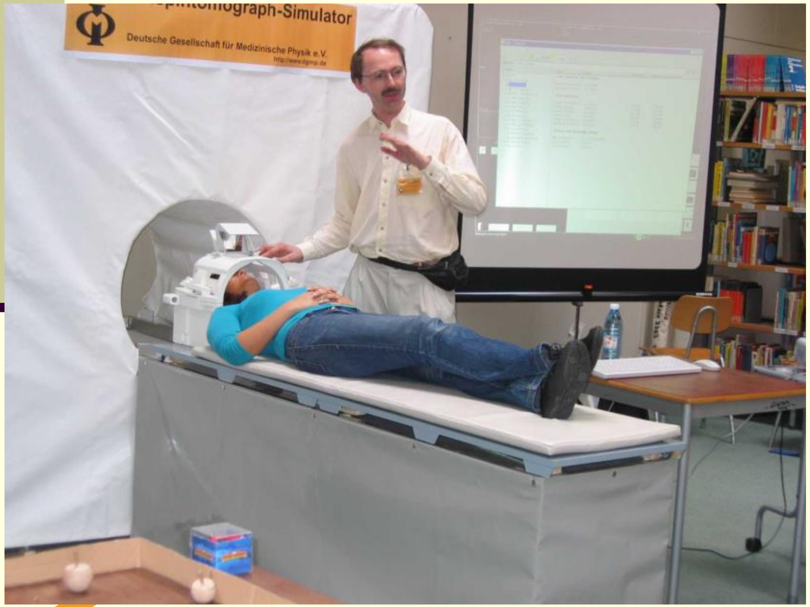
German Medical Physics Society‘s (DGMP) activity in MR

How to improve knowledge about the wonderful MR technique?

Take the MR to the people!

- MR-Simulator for PR-purpose
- Get young people interested in physics and specially in Medical Physics!!

The Future of MR is in the brain of young scientists!



Acknowledgements

- F. Schick, UKT Tübingen
- B. Pichler, UKT Tübingen
- K. Prüssmann, ETH Zürich
- R. Pohmann MPI for Biological Cybernetics, Tübingen
- W.Schlegel and L. Schad, DKFZ Heidelberg

- Siemens
- Philips
- ESAOTE
- Bruker Biospin

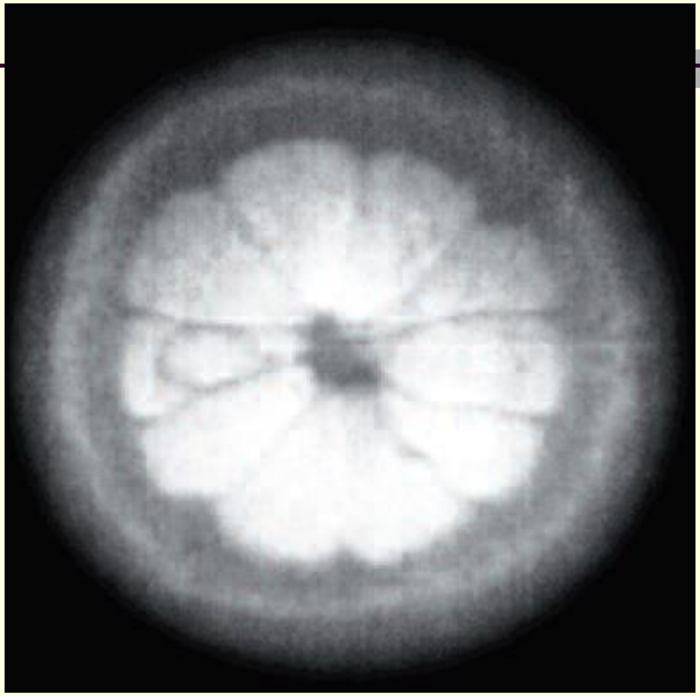
Eberhard-Karls-Universität
UKT
Universitätsklinikum Tübingen

medphys

Teşekkürler

Early MRI at Philips /NL

The Proton project team. Jürgen Heinzerling holds the special tool used to adjust the magnet coils in the shimming procedure.



The Nottingham Orange, 1978

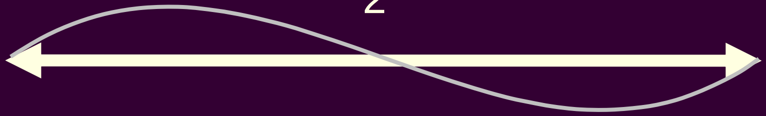
Eberhard-Karls-Universität UKT Universitätsklinikum Tübingen

From: MedicaMundi 50/1 2006/5

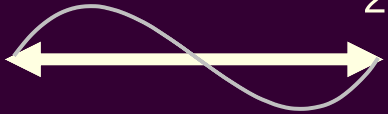
medphys

RF wavelength in water for different field strengths

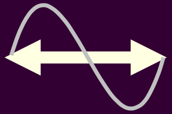
1.5 Tesla: λ_{H_2O} = 52 cm

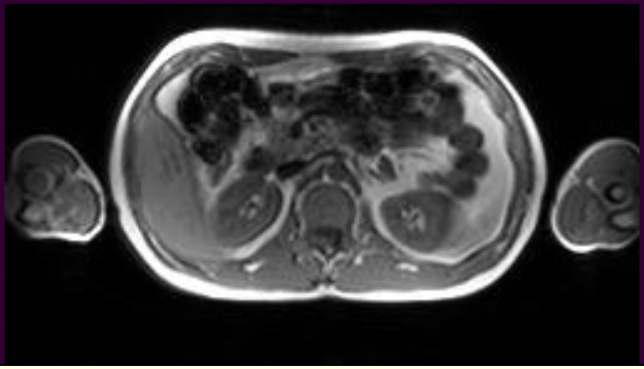


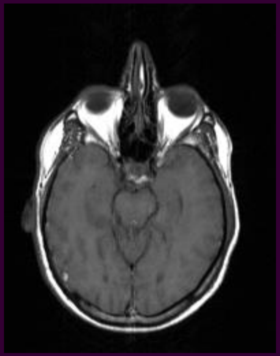
3.0 Tesla: λ_{H_2O} = 26 cm



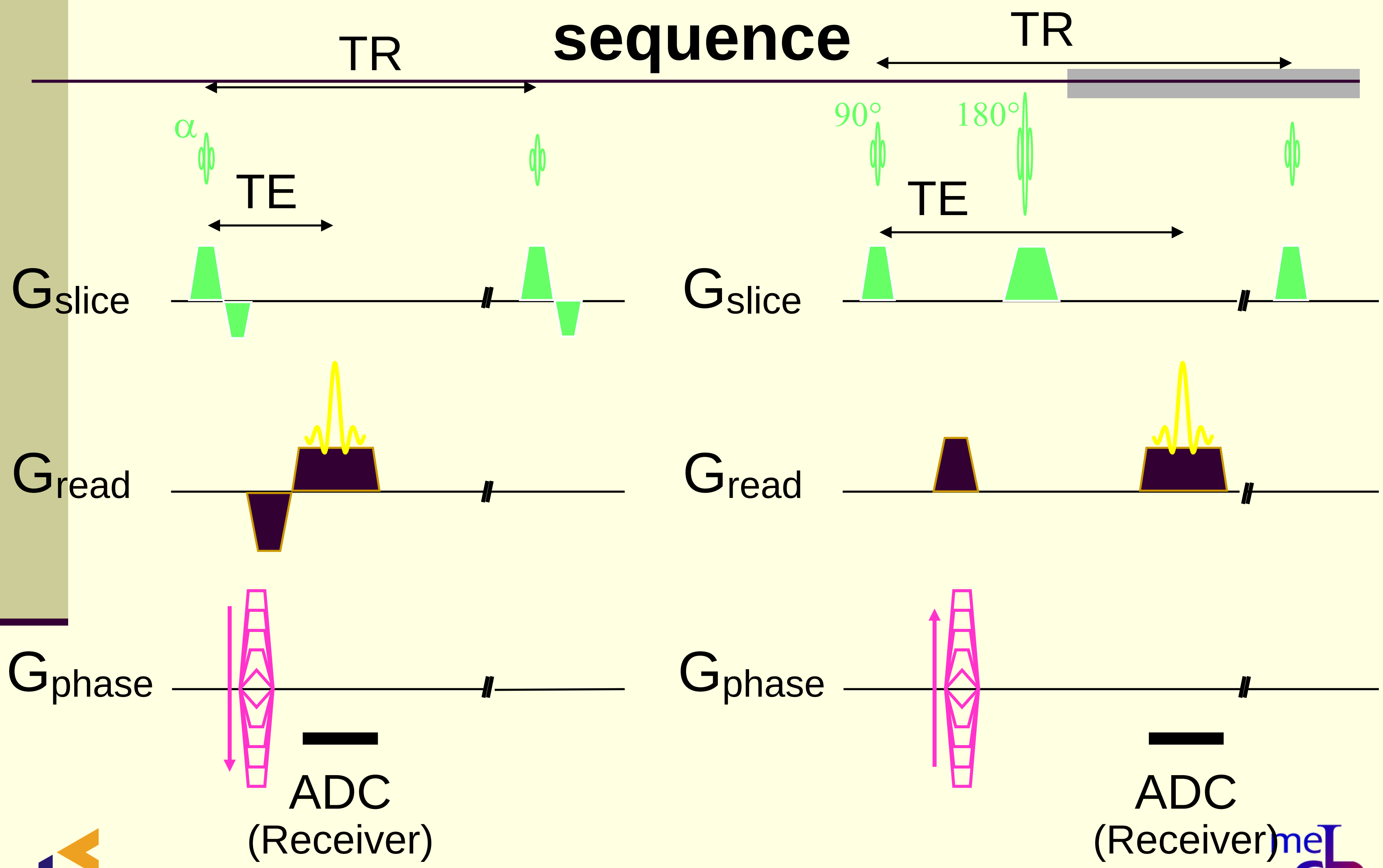
7.0 Tesla: λ_{H_2O} = 11 cm



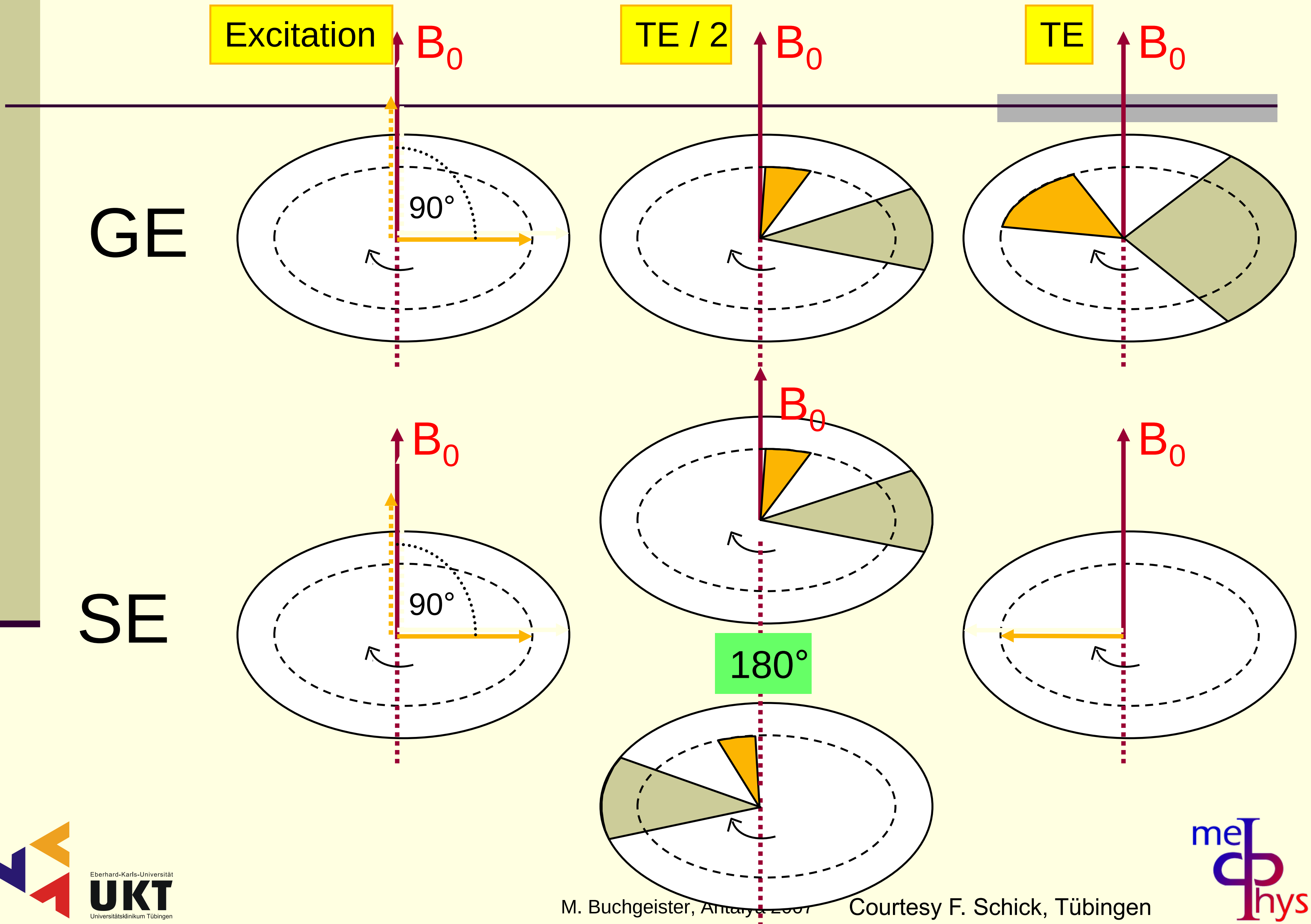




Gradient- and Spin echo- sequence



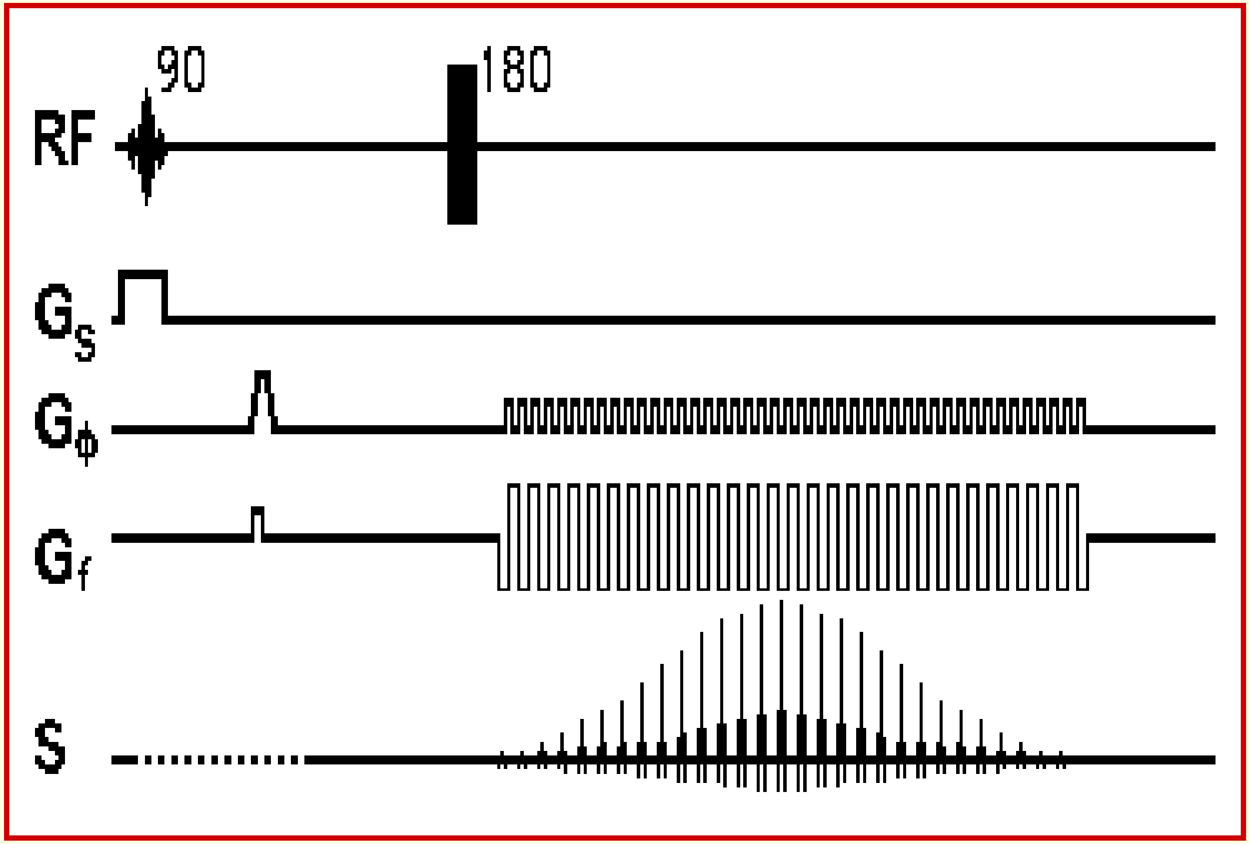
Dephasing of GE- and SE-sequences

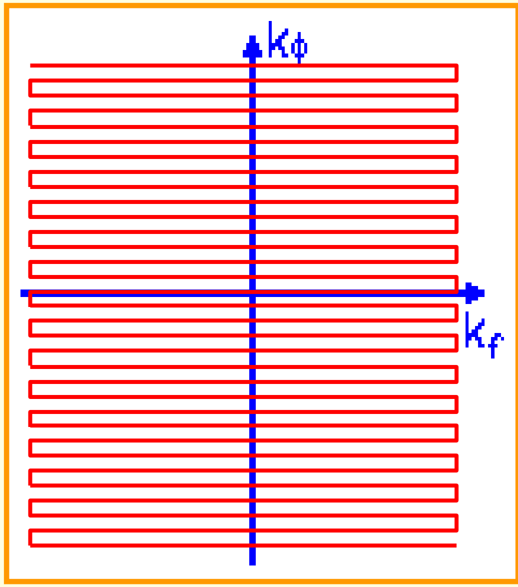


Courtesy F. Schick, Tübingen

Fast MR Imaging pulse sequences

Echo-Planar Imaging (EPI)





Courtesy F. Schick, Tübingen

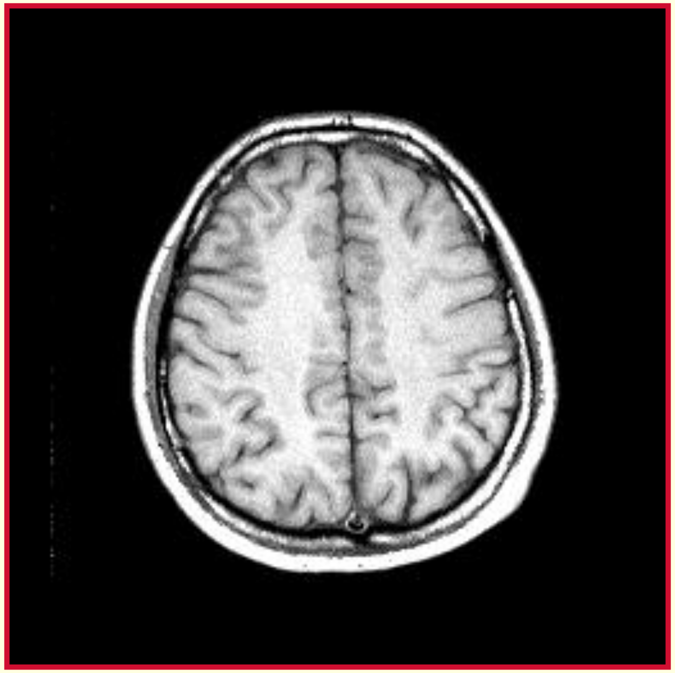
Fast MR Imaging pulse sequences

Echo-Planar Imaging (EPI)

- The whole k-space is filled after one excitation
- A fast oscillating gradient (in read-out direction) is switched on
- Phase encoding is done by short gradient pulses (blips)
- Signal measuring should be done in about T2* time (T2 relaxation, inhomogeneities)
- Read gradients have to be switched fast

Fast MR Imaging pulse sequences

Echo-Planar Imaging (EPI)



SE-Tomogram
TA > 1 min

EPI-Tomogram
TA = 0,3 s

Courtesy F. Schick, Tübingen

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Fast MR Imaging pulse sequences

Echo-Planar Imaging (EPI)

Advantages and Disadvantages:

+ Very short acquisition time (<< 1 s)

- High homogeneity of the field necessary due to T2*-effect (signal vanishes fast!)
- No 180°-pulses -> phase errors add up -> image errors
- Phase errors shifts fat signal several pixels in image -> fat saturation necessary