MRI basics

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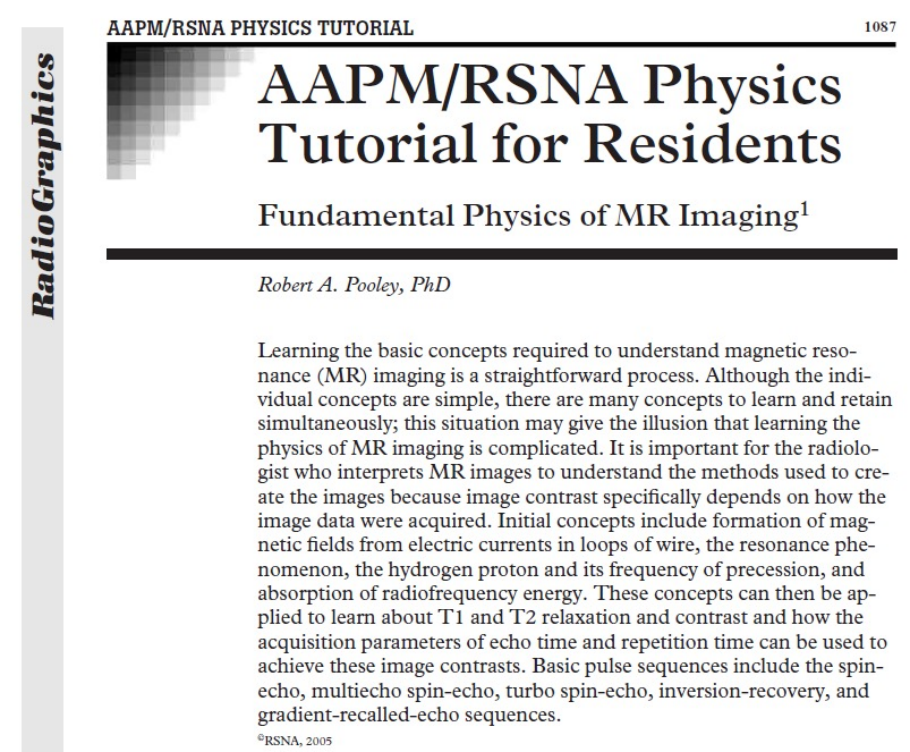
Recommendations for more information

www.imaios.com

www.mriquestions.com

www.revisemri.com

Homework for MRI lectures



RadioGraphics

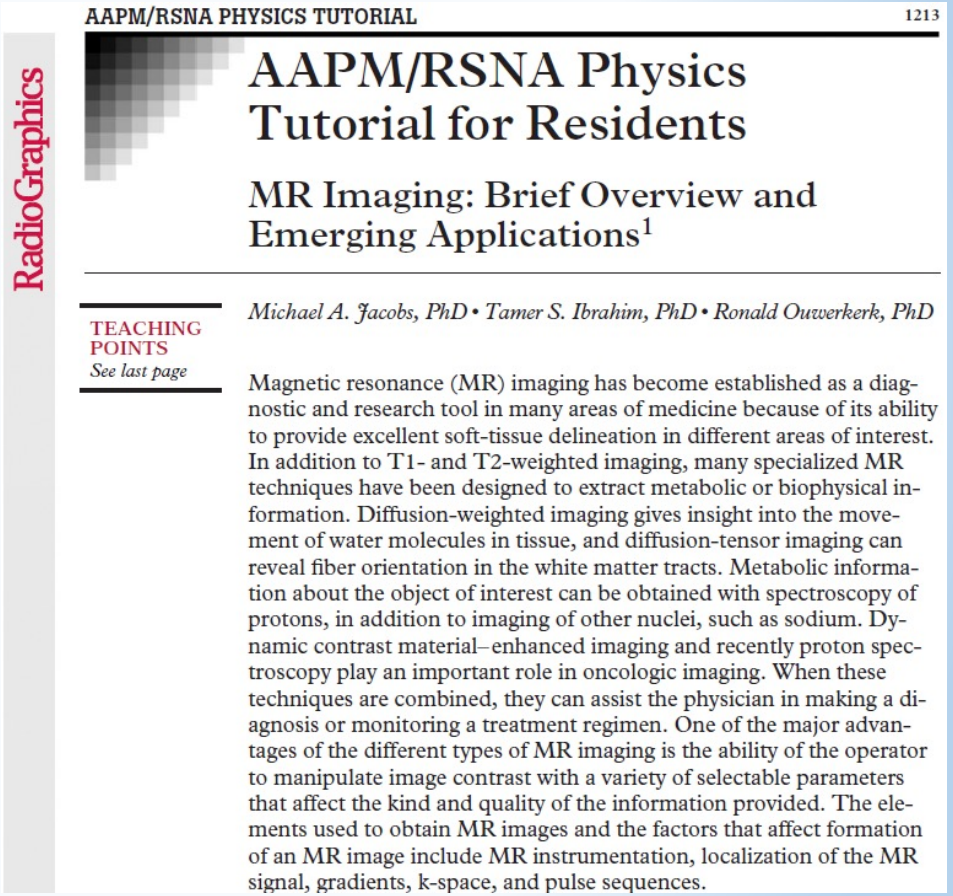
AAPM/RSNA PHYSICS TUTORIAL 1087

AAPM/RSNA Physics Tutorial for Residents

Fundamental Physics of MR Imaging[1]

Robert A. Pooley, PhD

Learning the basic concepts required to understand magnetic resonance (MR) imaging is a straightforward process. Although the individual concepts are simple, there are many concepts to learn and retain simultaneously; this situation may give the illusion that learning the physics of MR imaging is complicated. It is important for the radiologist who interprets MR images to understand the methods used to create the images because image contrast specifically depends on how the image data were acquired. Initial concepts include formation of magnetic fields from electric currents in loops of wire, the resonance phenomenon, the hydrogen proton and its frequency of precession, and absorption of radiofrequency energy. These concepts can then be applied to learn about T1 and T2 relaxation and contrast and how the acquisition parameters of echo time and repetition time can be used to achieve these image contrasts. Basic pulse sequences include the spin-echo, multiecho spin-echo, turbo spin-echo, inversion-recovery, and gradient-recalled-echo sequences.



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AAPM/RSNA PHYSICS TUTORIAL 1213

AAPM/RSNA Physics Tutorial for Residents

MR Imaging: Brief Overview and Emerging Applications[1]

TEACHING POINTS
See last page

Michael A. Jacobs, PhD • Tamer S. Ibrahim, PhD • Ronald Ouwerkerk, PhD

Magnetic resonance (MR) imaging has become established as a diagnostic and research tool in many areas of medicine because of its ability to provide excellent soft-tissue delineation in different areas of interest. In addition to T1- and T2-weighted imaging, many specialized MR techniques have been designed to extract metabolic or biophysical information. Diffusion-weighted imaging gives insight into the movement of water molecules in tissue, and diffusion-tensor imaging can reveal fiber orientation in the white matter tracts. Metabolic information about the object of interest can be obtained with spectroscopy of protons, in addition to imaging of other nuclei, such as sodium. Dynamic contrast material–enhanced imaging and recently proton spectroscopy play an important role in oncologic imaging. When these techniques are combined, they can assist the physician in making a diagnosis or monitoring a treatment regimen. One of the major advantages of the different types of MR imaging is the ability of the operator to manipulate image contrast with a variety of selectable parameters that affect the kind and quality of the information provided. The elements used to obtain MR images and the factors that affect formation of an MR image include MR instrumentation, localization of the MR signal, gradients, k-space, and pulse sequences.

Magnetic Resonance Imaging

Principles of MRI (very short version)

- Static magnetic field creates "*net magnetization*" of nuclear spins.
- RF Field is used to "*excite spins*" and to alter net magnetization to observable signal
- Gradient fields are used to "*encode*" information about spin location to signal
- Mathematical method called *Fast Fourier Transformation* (*FFT*) is used to create images from acquired signals

Principles of MRI (very short version)

Appearance of the resulting image depends from the use of RF-pulses and gradients during the imaging *sequence*.

…meaning that image *resolution* and *contrast* depend from the used *scanning parameters*.

Principles of MRI (very short version)

In order to understand how we get magnetic resonance images (MR images) to look just as we want them to look like (*optimal resolution and contrast*), we need to understand underlying physics..,

…which, unfortunately, is not easy.

Why Magnetic Resonance Imaging

- complete absence of ionizing radiation
- variety of contrast mechanisms

X-ray CT

the linear X-ray attenuation coefficient

1H MRI

Proton density, PD
longitudinal relaxation time, *T*1
transverse relaxation time, *T*2

Flow sensitivity
Perfusion
Diffusion
Tissue's magnetic susceptibility
Spectroscopy

Nuclear Magnetic Resonance

* **N**uclear **M**agnetic **R**esonance (NMR), 1946 (Bloch & Purcell)

  "NMR is the physical basis of MRI and MRS"

* **Spin Echo Technique** (NMR & MRI), 1952 (Hahn)

* **M**agnetic **R**esonance **I**maging (MRI), 1973 P. Lauterbur

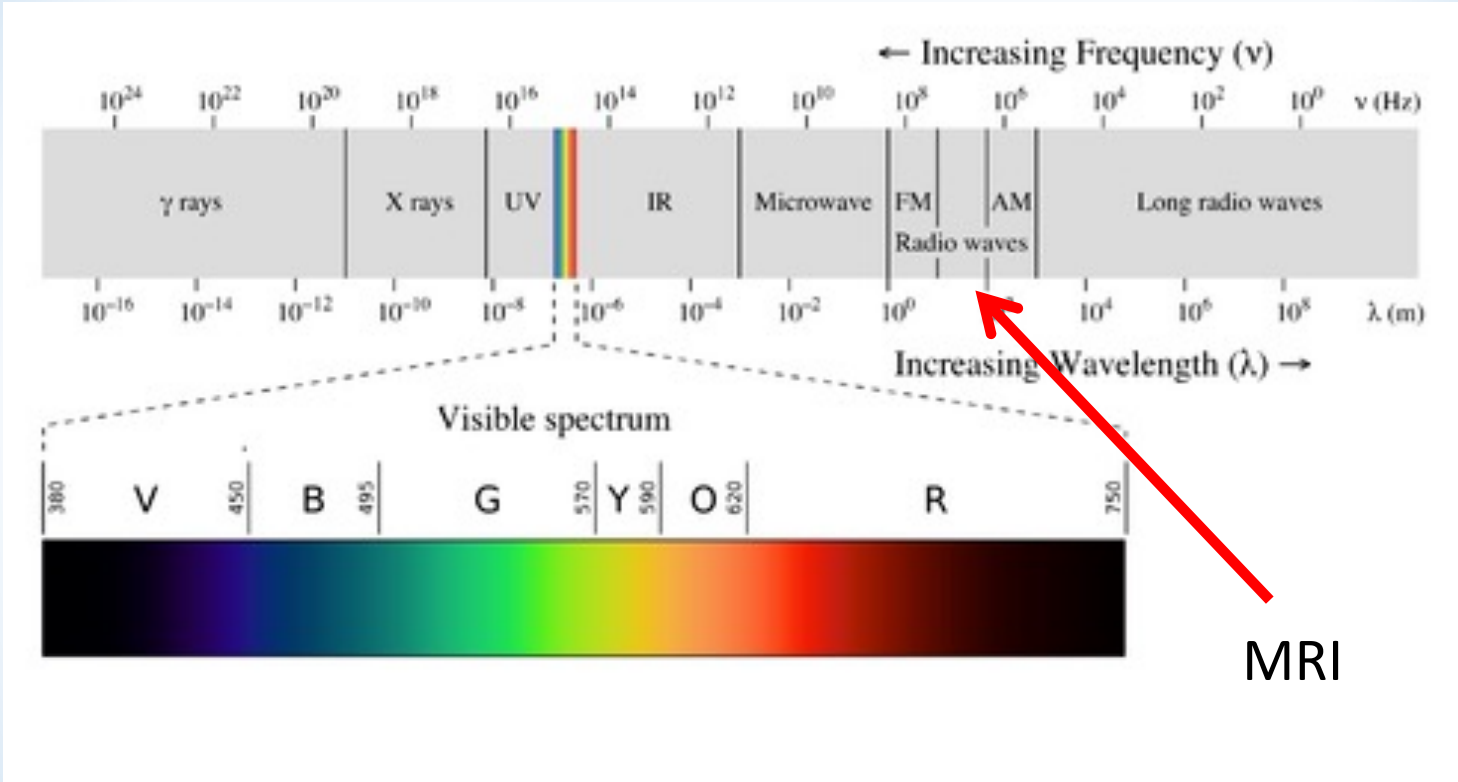
* **MRI research in Turku** (MRI), 1980, (First scanner 1984)

* The 2003 Nobel Prize to P. Lauterbur and Sir P. Mansfield

Nuclear Magnetic Resonance

Nuclear Magnetic Resonance can be described as a phenomenon taking place when atomic nuclei in a static magnetic field absorb energy from a radiofrequency (RF) field of certain frequency characteristic to the nuclei.

Energies in MRI



Nuclear Magnetic Resonance

Spin

- Every nucleus has a property called spin.

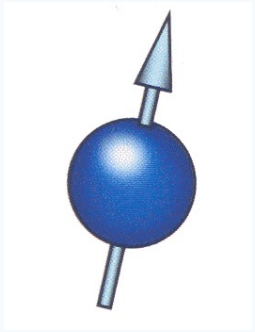
- It is an intrinsic property of the particle itself, just as mass or charge of the particle.

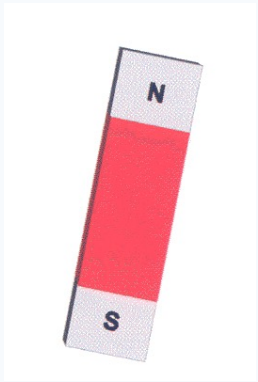
- Spin can be considered as a form of angular momentum, although it is not produced by the rotation of the particle.

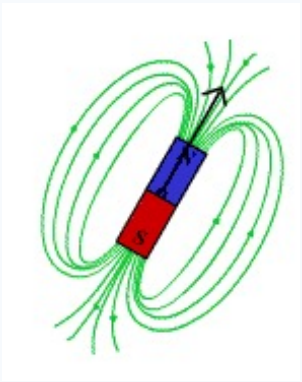
Nuclear Magnetic Resonance

Magnetic moment

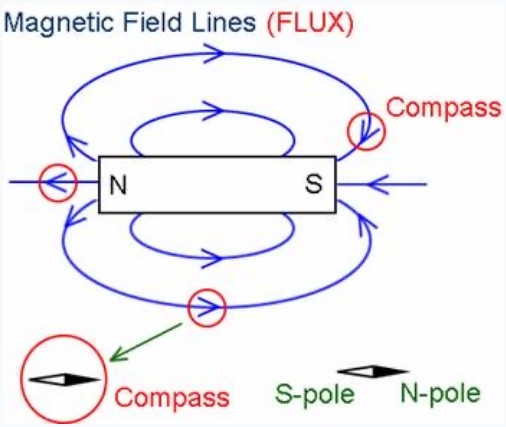
Spin and magnetism are very closely linked. In addition to spin, nuclei with nonzero spin ($I > 0$) also possess intrinsic nuclear magnetism called **magnetic moment**.







Compass needle in magnetic field



Macroscopic magnetization M_z

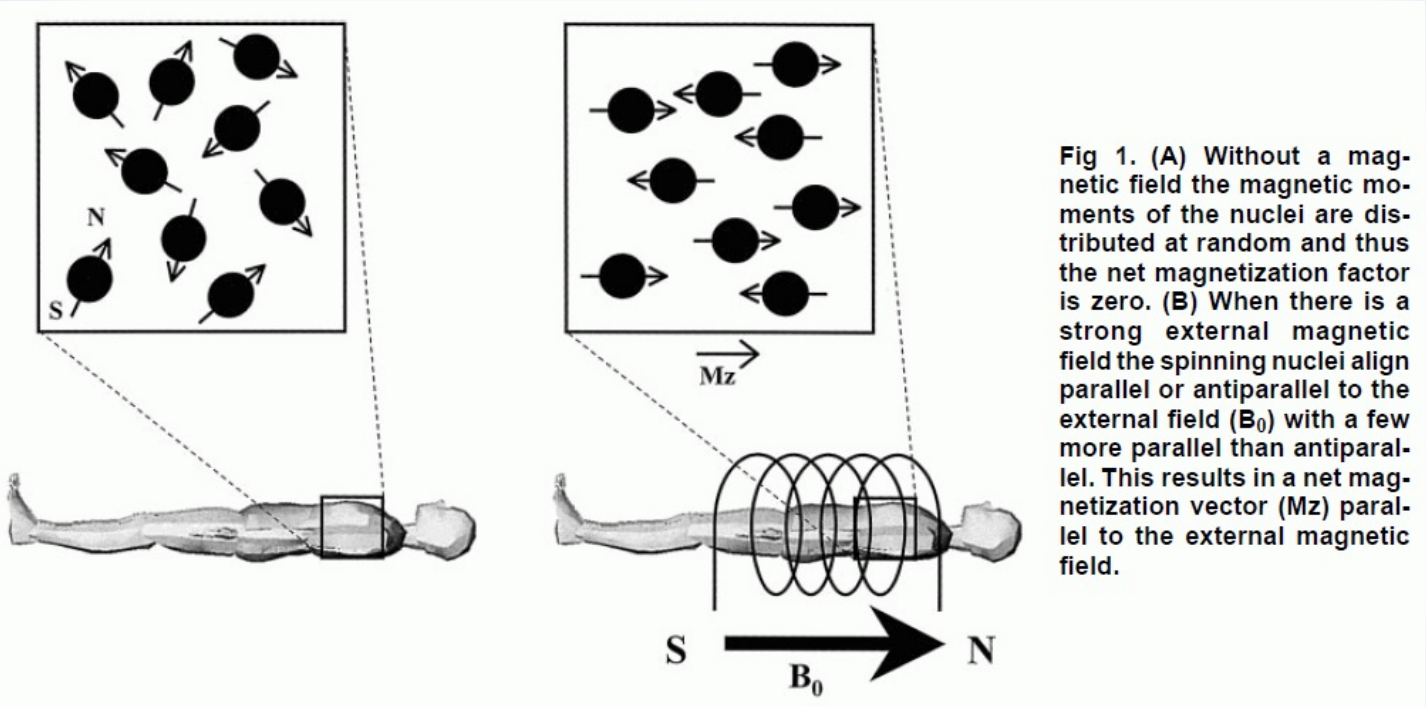
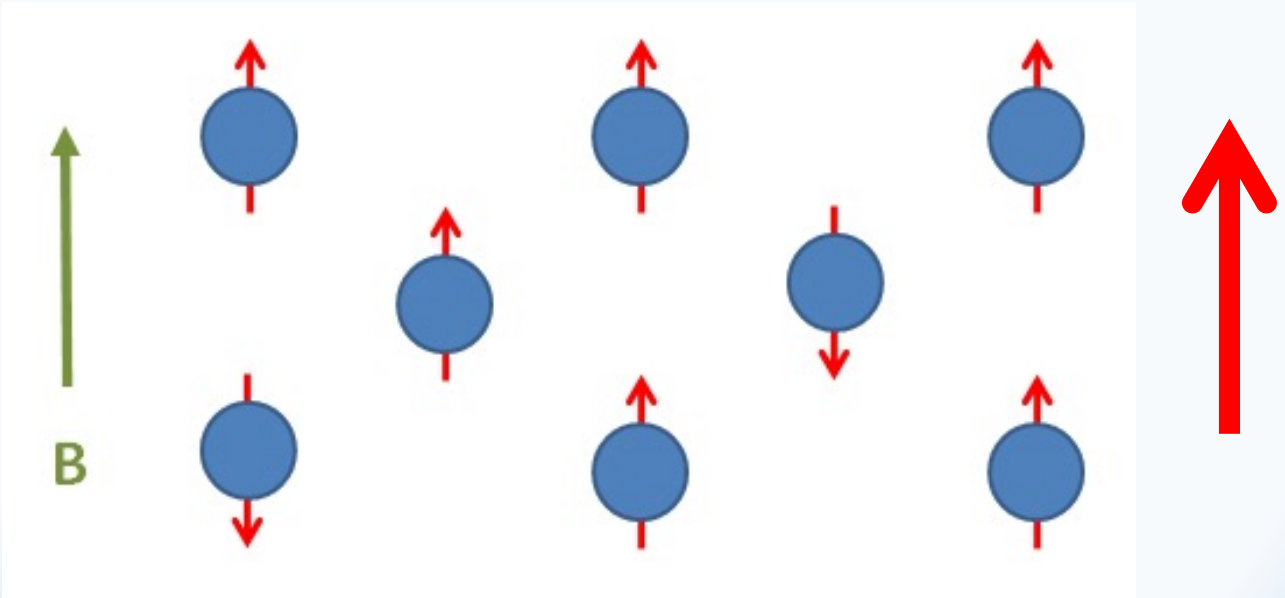


Fig 1. (A) Without a magnetic field the magnetic moments of the nuclei are distributed at random and thus the net magnetization factor is zero. (B) When there is a strong external magnetic field the spinning nuclei align parallel or antiparallel to the external field (B_0) with a few more parallel than antiparallel. This results in a net magnetization vector (Mz) parallel to the external magnetic field.

Magnitude and direction of magnetic moment can be described by a vector, so you can calculate a sum of magnetic moments of several nuclei. The result is called **NET MAGNETIZATION** (VECTOR) or **MACROSCOPIC MAGNETIZATION**.



Spin and magnetic field
(quantum mechanical model)

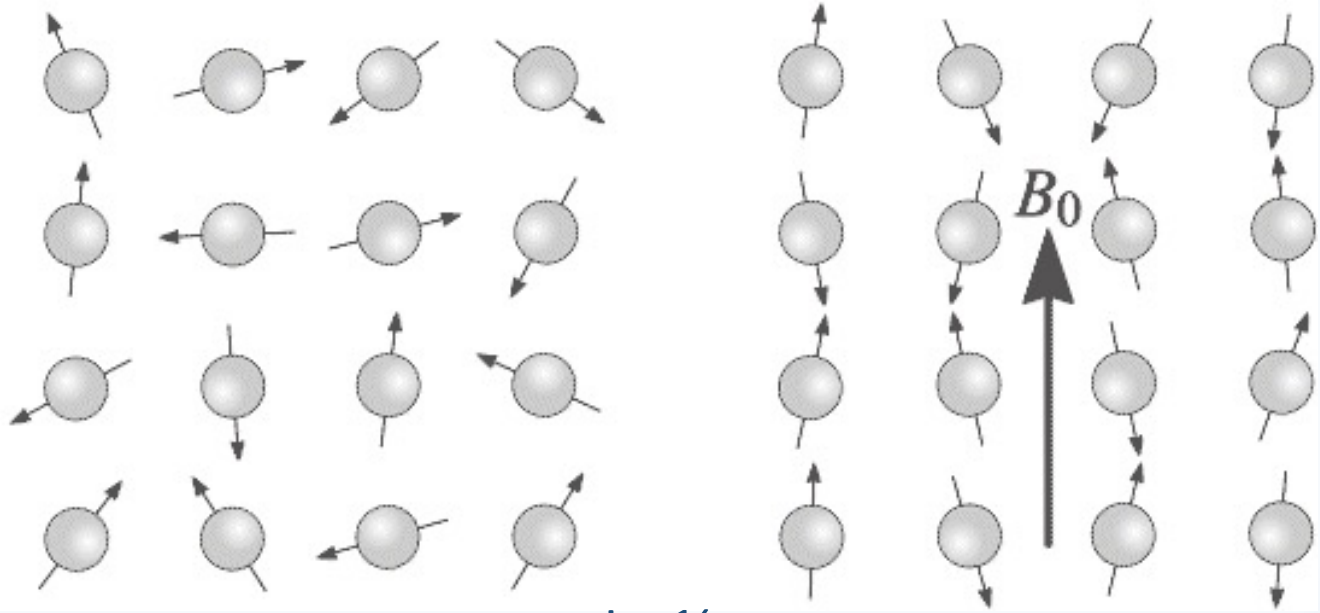
Zeeman interaction

When external magnetic field $\mathbf{B}_0$ is applied, an interaction called the Zeeman interaction exists between the magnetic moment of a nucleus and $\mathbf{B}_0$.

As a result energy levels of the nucleus are quantized as well and particles with spin I have $2I + 1$ sublevels with different energies.

Spin and magnetic field (quantum mechanical model)

Two energy states of ^{1}H spins in magnetic field



I = ½

→ 2* ½ + 1 = 2 sublevels

Spin and magnetic field
(quantum mechanical model)

When I > ½, it is more complicated

I = 1

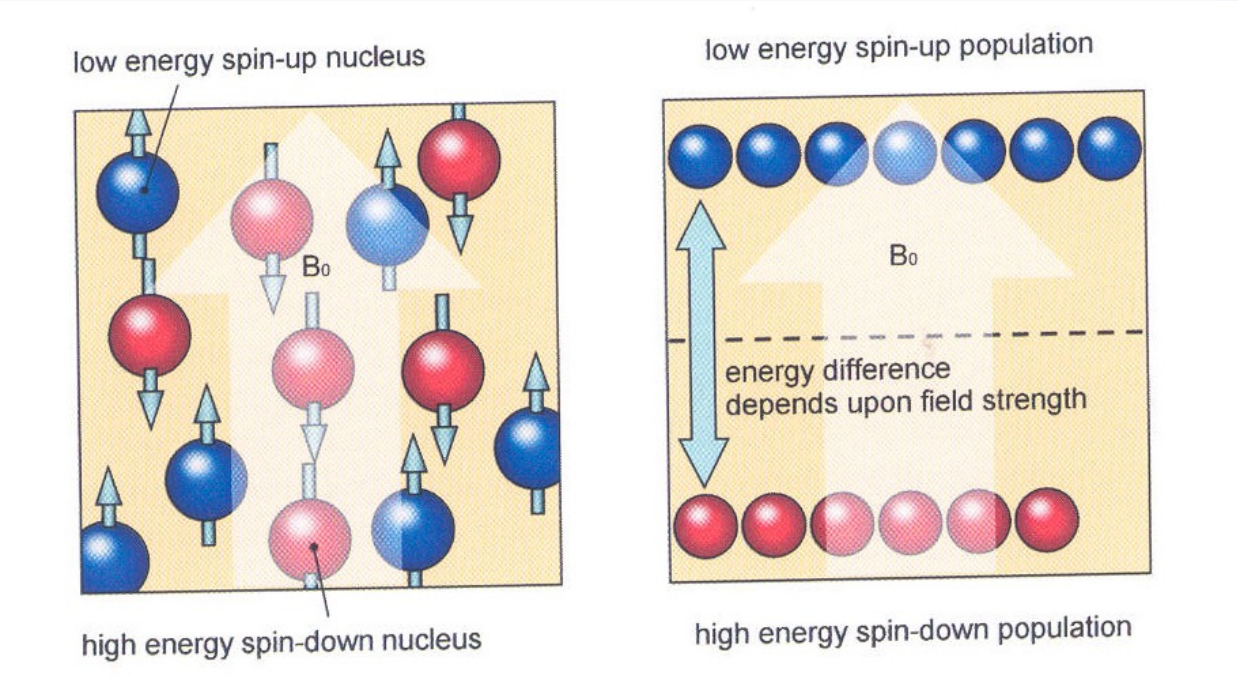
→ 2* 1 + 1 = 3 sublevels

I = 3/2

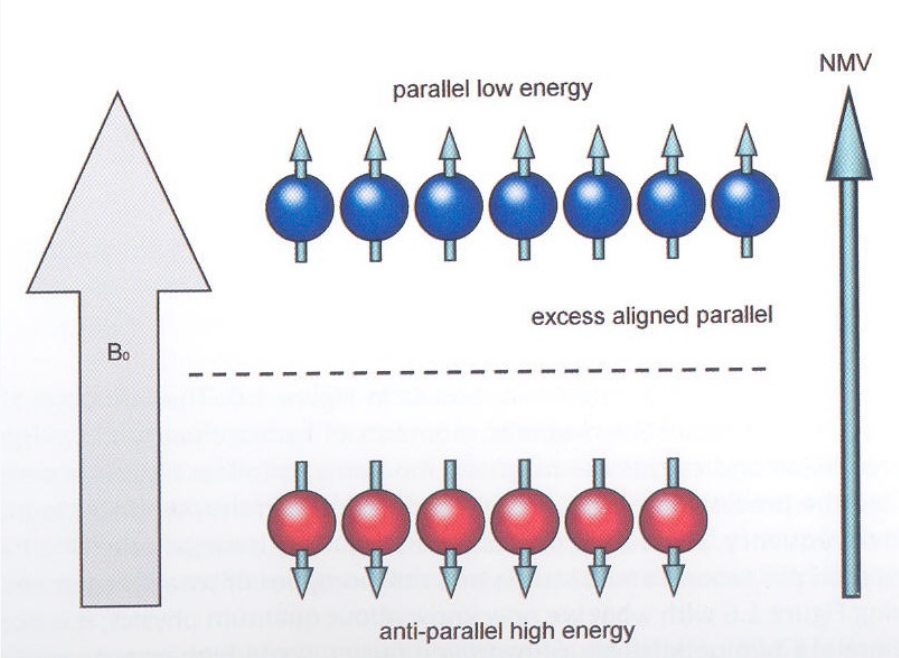
→ 2* 3/2 + 1 = 4 sublevels

In the case of MRI, it is enough to concentrate on I = ½ nuclei (and in practice to ^{1}H).

Spin and magnetic field (quantum mechanical model)



Spin and magnetic field (quantum mechanical model)



According to **Boltzmann statistics**, the low energy level will have a higher population than the higher energy level, so that **thermal equilibrium** is reached

Nuclear spin polarization

Relative populations of adjacent energy levels in thermal equilibrium are given by the Boltzmann distribution

$$\frac{n_{m+1}}{n_m} = \exp\left(-\frac{\gamma\hbar B_0}{kT}\right) \qquad (1)$$

h = Planck constant
k = Boltzmann constant

Nuclear spin polarization φ is defined as

$$\varphi = \left|\frac{n_m - n_{m+1}}{n_m + n_{m+1}}\right| \qquad (2)$$

Using equation (1) this can be expressed as

$$\varphi \approx \left|\frac{\gamma\hbar B_0}{2kT}\right|, \qquad |\gamma|\hbar B_0 << kT \qquad (3)$$

Nuclear spin polarization

^{1}H nucleus nuclear spin polarization in 3T magnet (T = 295 K)

$$\varphi \approx \left| \frac{\gamma \hbar B_0}{2kT} \right| \approx 10 \text{ ppm} = 0{,}001\%$$

Nuclear spin polarization enhancement:

* Stronger magnetic field
* Lower temperature

Nuclear spin polarization

1 mm^3 of water contains about $6{,}7 \cdot 10^{22}$ hydrogen nuclei.

Despite the small population difference of adjacent sublevels, this huge number means that there are enough small magnetic moments to create observable **net magnetization**!

Diagnostically relevant magnetic nuclei for MRI

Table 1-1. Magnetic resonance properties of some diagnostically relevant nuclei

	Nucleus		Relative abundance (%)	Relative sensitivity*	Magnetogyric ratio (MHz/T)
MRI MRS	^{1}H	I = 1/2	99.98	1	42.58
	^{2}H		0.015	9.65×10^{-3}	6.53
	^{13}C		1.11	0.016	10.71
	^{19}F		100	0.830	40.05
	^{23}Na		100	0.093	11.26
MRS	^{31}P	I = 1/2	100	6.6×10^{-2}	17.23
	^{39}K		93.1	5.08×10^{-4}	1.99

*At constant field for equal number of nuclei

Nuclear Magnetic Resonance

In order to fully understand NMR phenomenon a quantum mechanical approach is required…

Fortunately, in order to understand NMR phenomenon on a level that we need, a classical approach is enough!

Nuclear Magnetic Resonance Imaging

During the early years of MRI word “Nuclear” was considered scary, so new imaging method was called just Magnetic Resonance Imaging (MRI).

In Finnish language even the word Resonance was omitted.

From the physical point of view, both should be mentioned…

Nuclear Magnetic Resonance Imaging

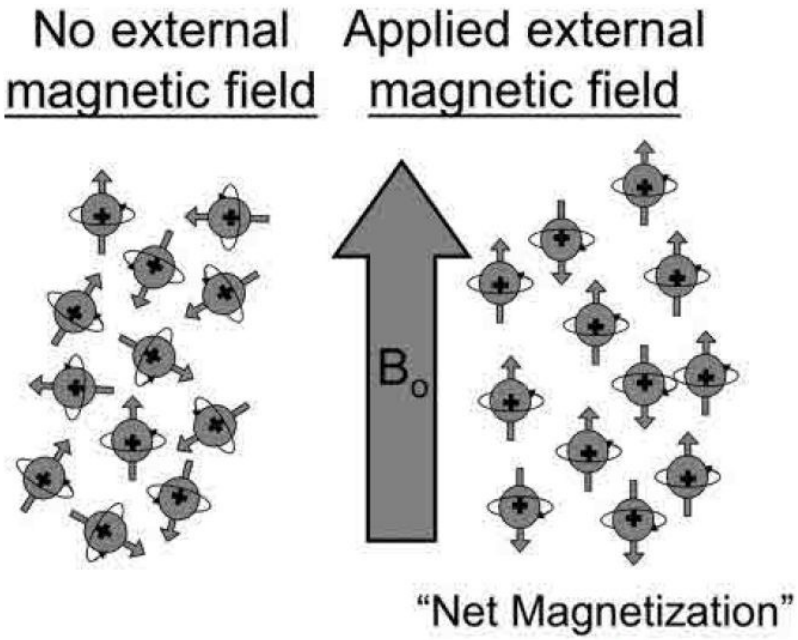
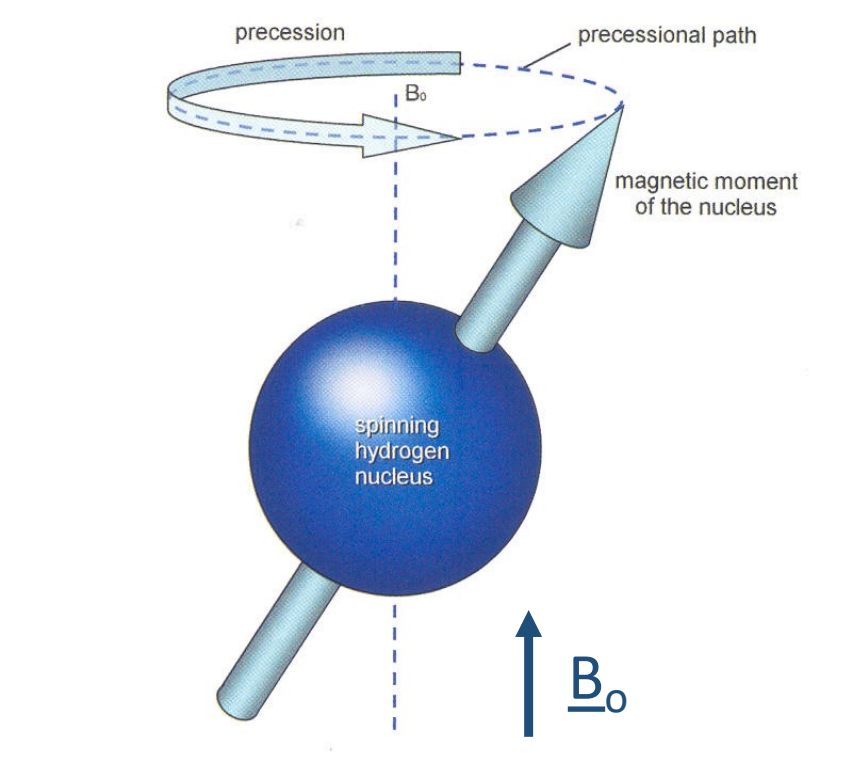


Figure 4. Alignment of protons with the B_0 field. With no external magnetic field, hydrogen protons *(+)* are oriented randomly. When the protons are placed in a strong magnetic field *(B_0)*, a net magnetization will be produced parallel to the main magnetic field.

Precession motion of the magnetic moment of nucleus in magnetic field



Z-axis is always considered to be in the direction of $\underline{B_0}$-field

Precession frequency (Larmor frequency) of protons is proportional to field strength intensity.

ω = Larmor frequency

$$\omega = \gamma \cdot B_0$$

^1H: $\gamma = 42{,}58$ MHz/T

Excitation

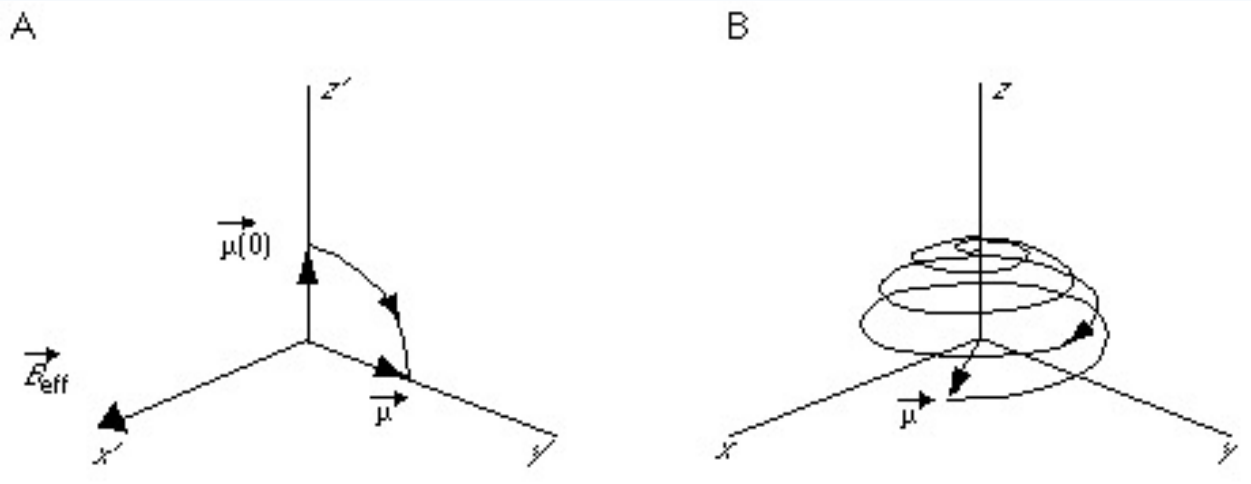
Exchange of energy between two systems at a specific frequency is called **resonance**. Magnetic resonance corresponds to the energetic interaction between spins and electromagnetic radiofrequency (RF).

Only protons that spin with the same frequency as the electromagnetic RF pulse will respond to that RF pulse. There is a modification of spin equilibrium and absorption of electromagnetic energy by atomic nuclei, which is called **excitation**.

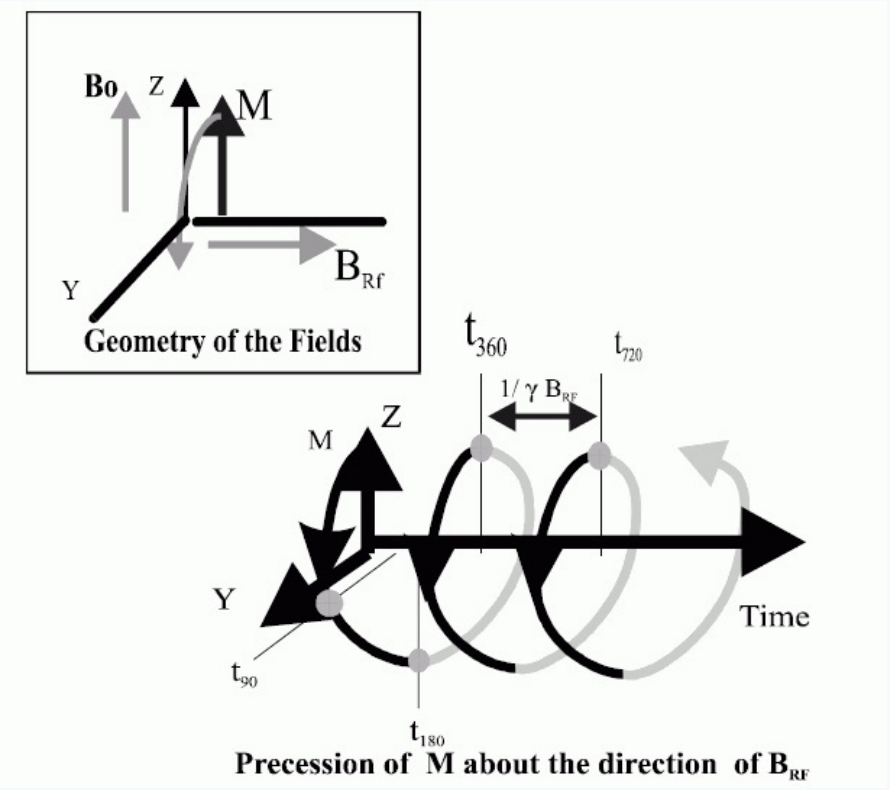
Rotating frame of reference

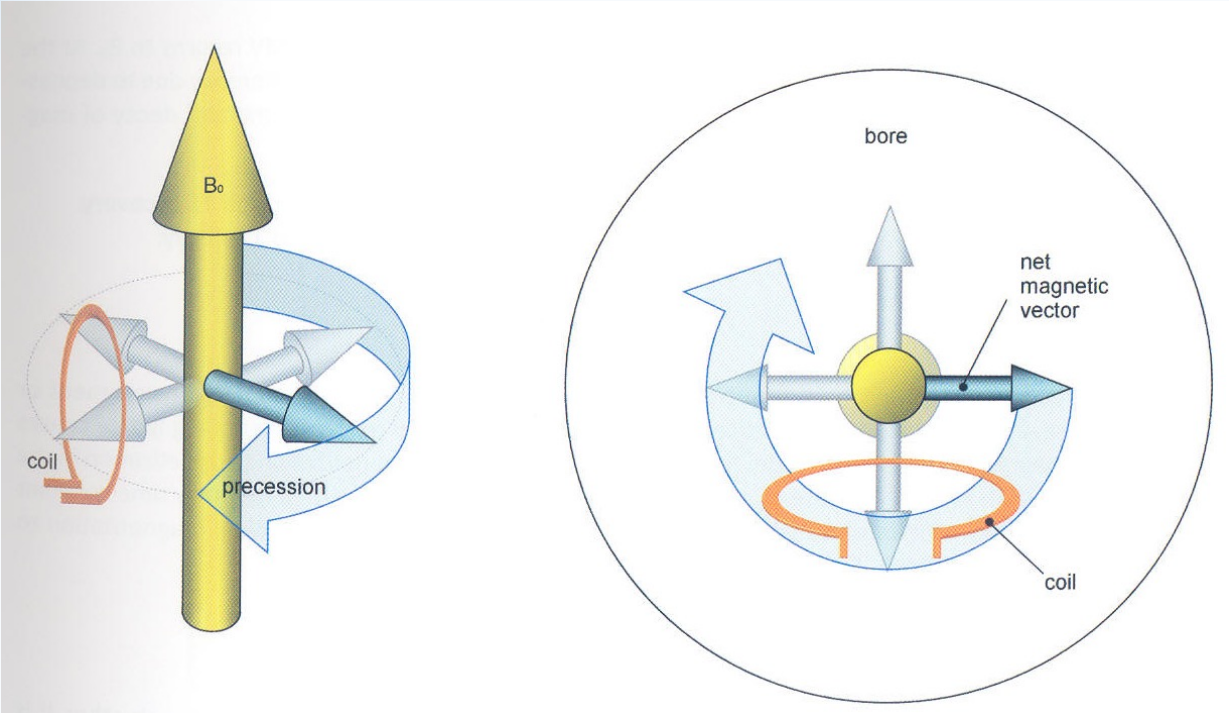
A)In a rotating frame of reference, the net magnetization vector tips down during excitation.

B) The exact consequence of the RF pulse on the macroscopic net magnetization vector is a spiral movement down to the XY plane.



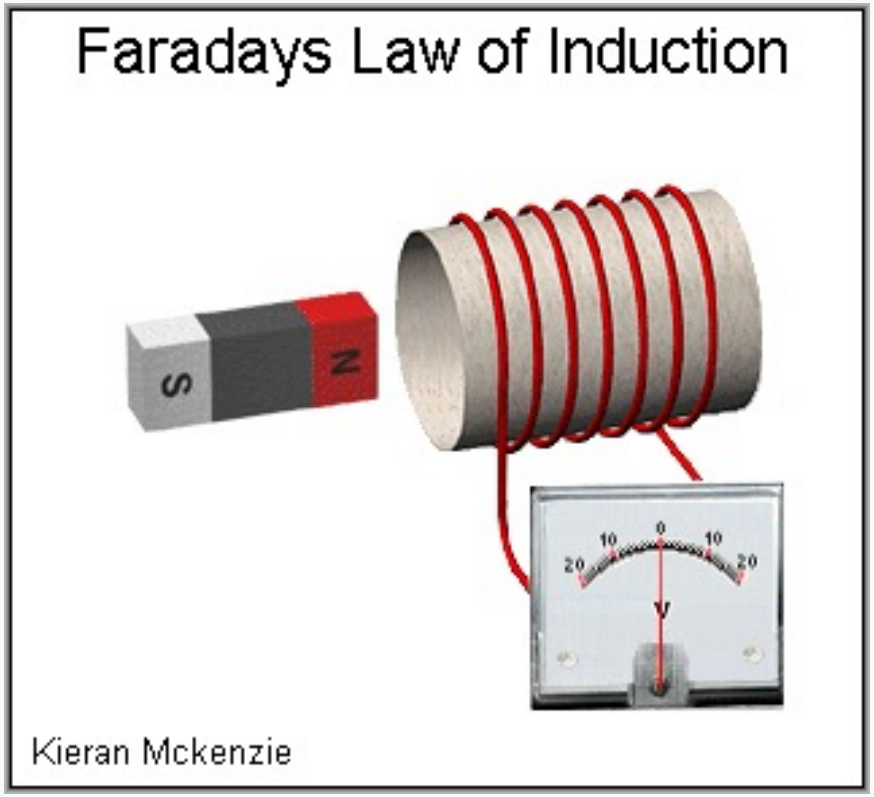
Effect of oscillating (Larmor frequency) RF-field on *M*
(rotating frame of reference)





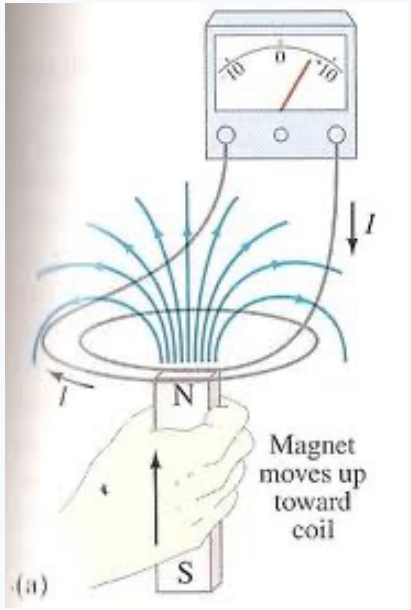
Immediately after 90 degree RF pulse all magnetic moments (spins) are in same phase, so Net Magnetization vector is completely in xy-plane.
Magnetic moments continue their precession motion and if there is a coil against xy-plane, there will be fluctuating magnetic field inside of it…

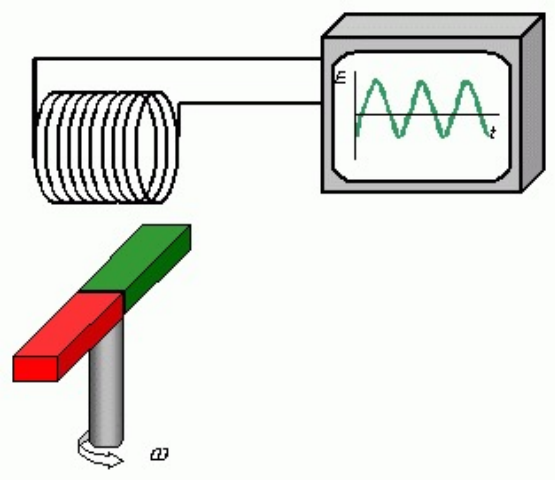
Electromagnetic induction



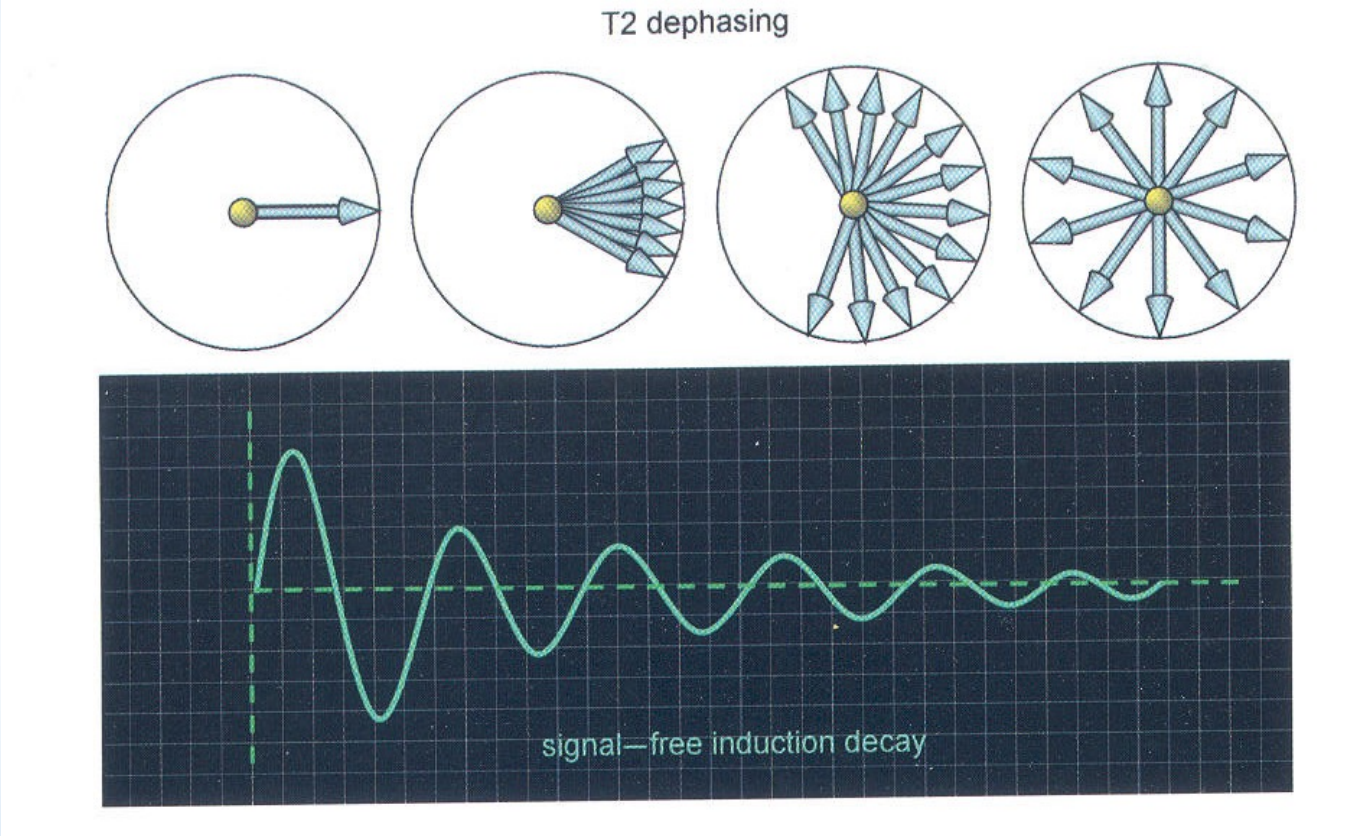
Receiver coils

Electromagnetic induction



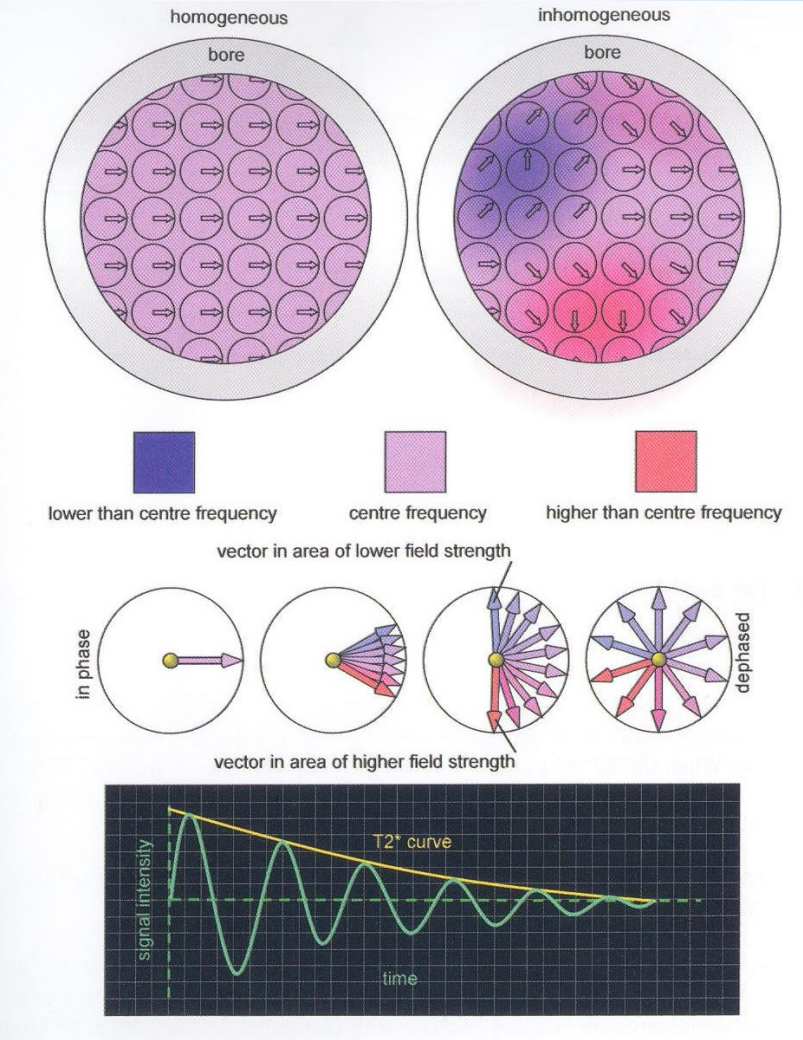


Free Induction Decay -signal

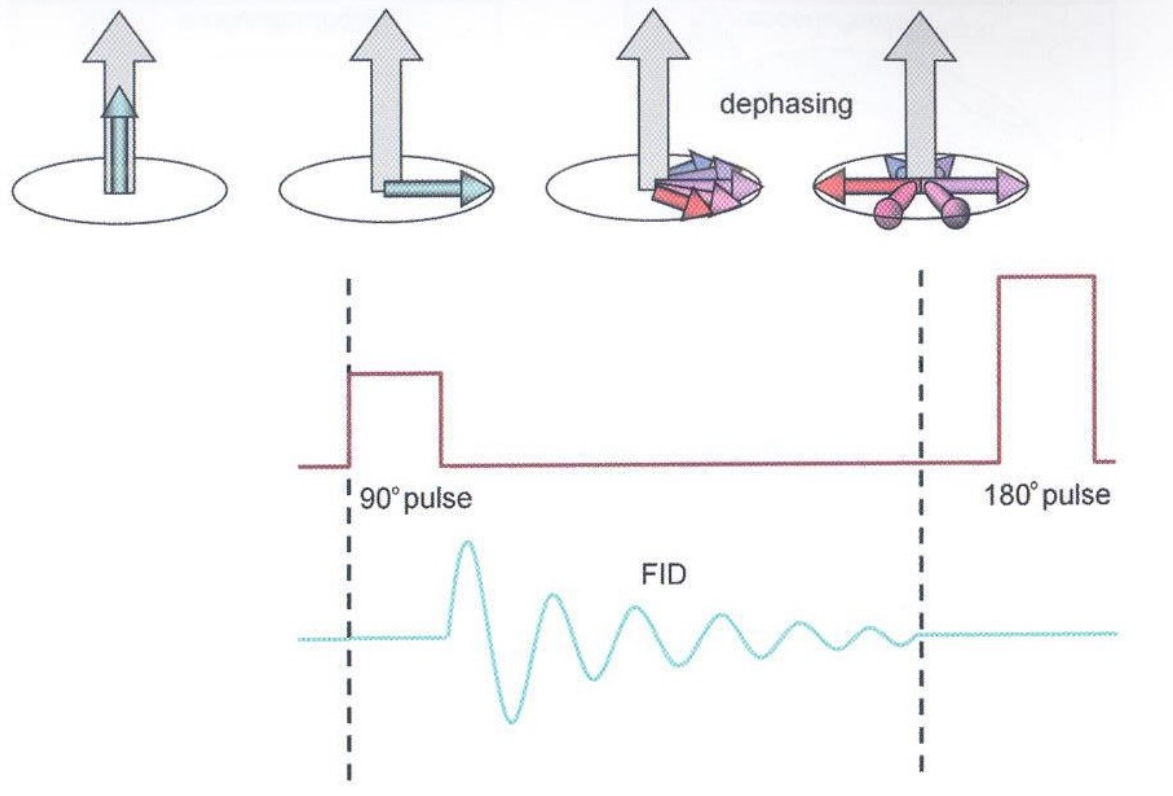


T2* relaxation

In an idealized system, all nuclei in a given chemical environment, in a magnetic field, precess with the same frequency. However, in real systems, there are minor differences in chemical environment which can lead to a distribution of resonance frequencies around the ideal. Over time, this distribution can lead to a dispersion of the tight distribution of magnetic spin vectors, and loss of signal (Free Induction Decay).



T2* relaxation

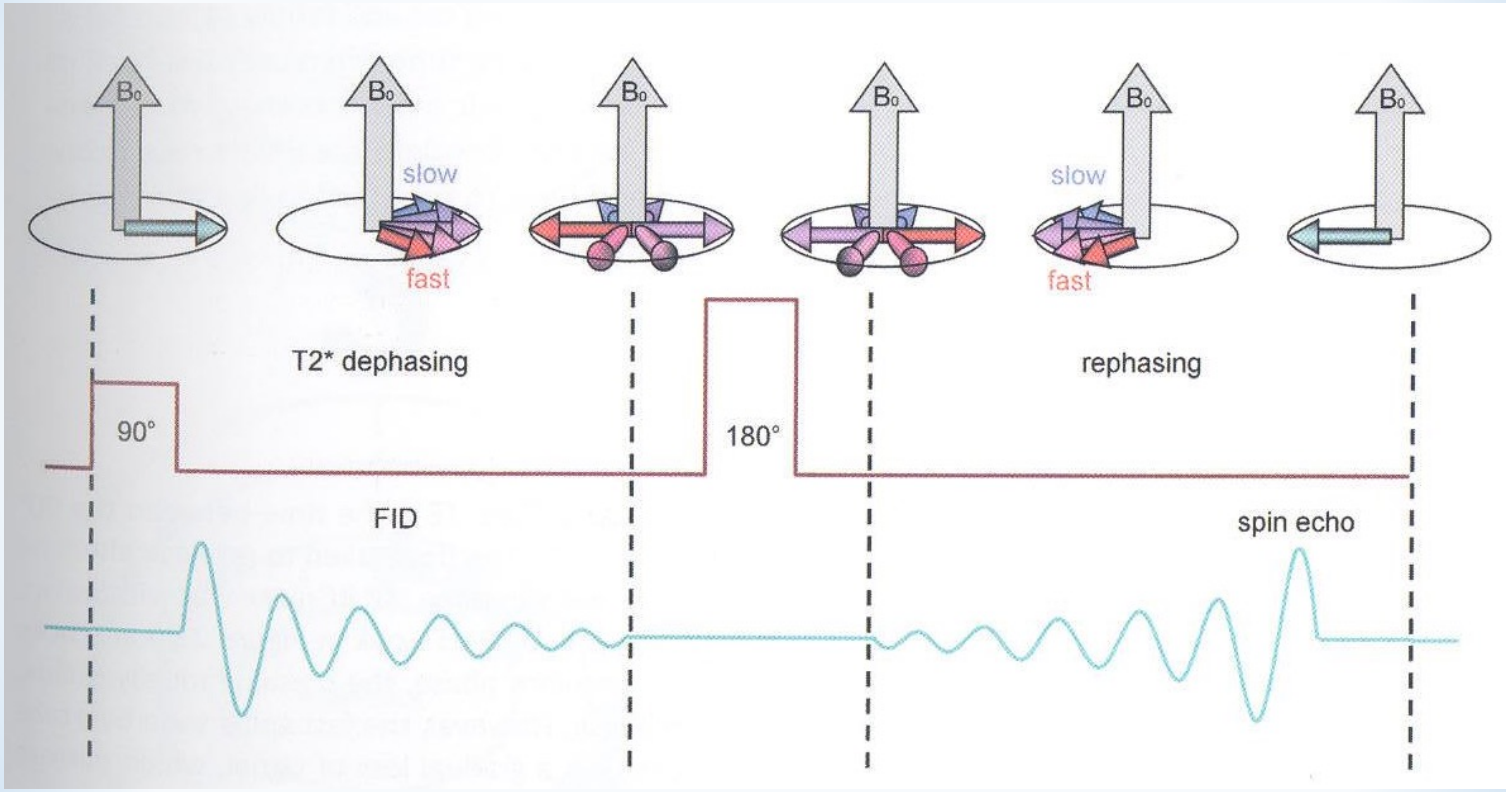


T2* relaxation

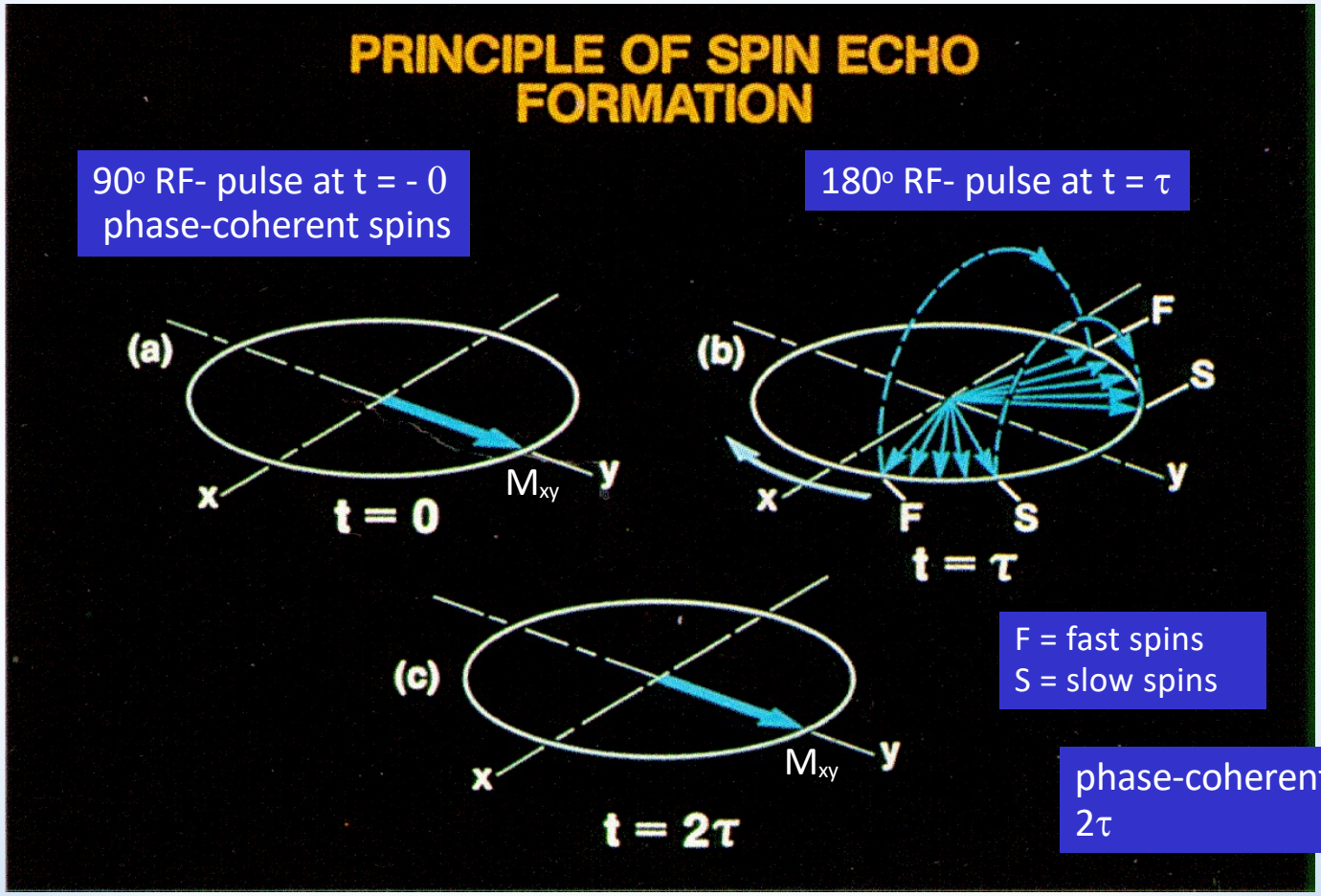
Decoherence because of magnetic field inhomogeneity is not a true "relaxation" process; it is not random, but dependent on the location of the molecule in the magnet.

For molecules that aren't moving, the deviation from ideal relaxation is consistent over time, and the signal can be recovered by performing a spin echo experiment.

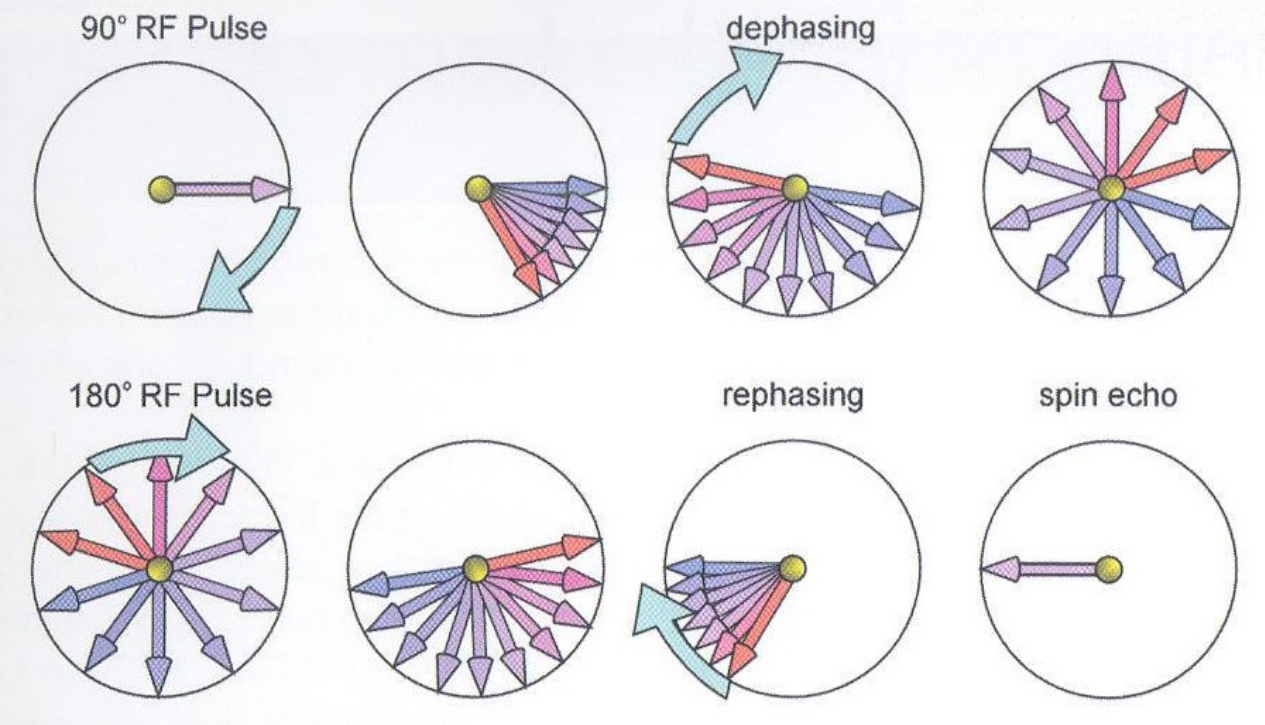
Spin-Echo sequence



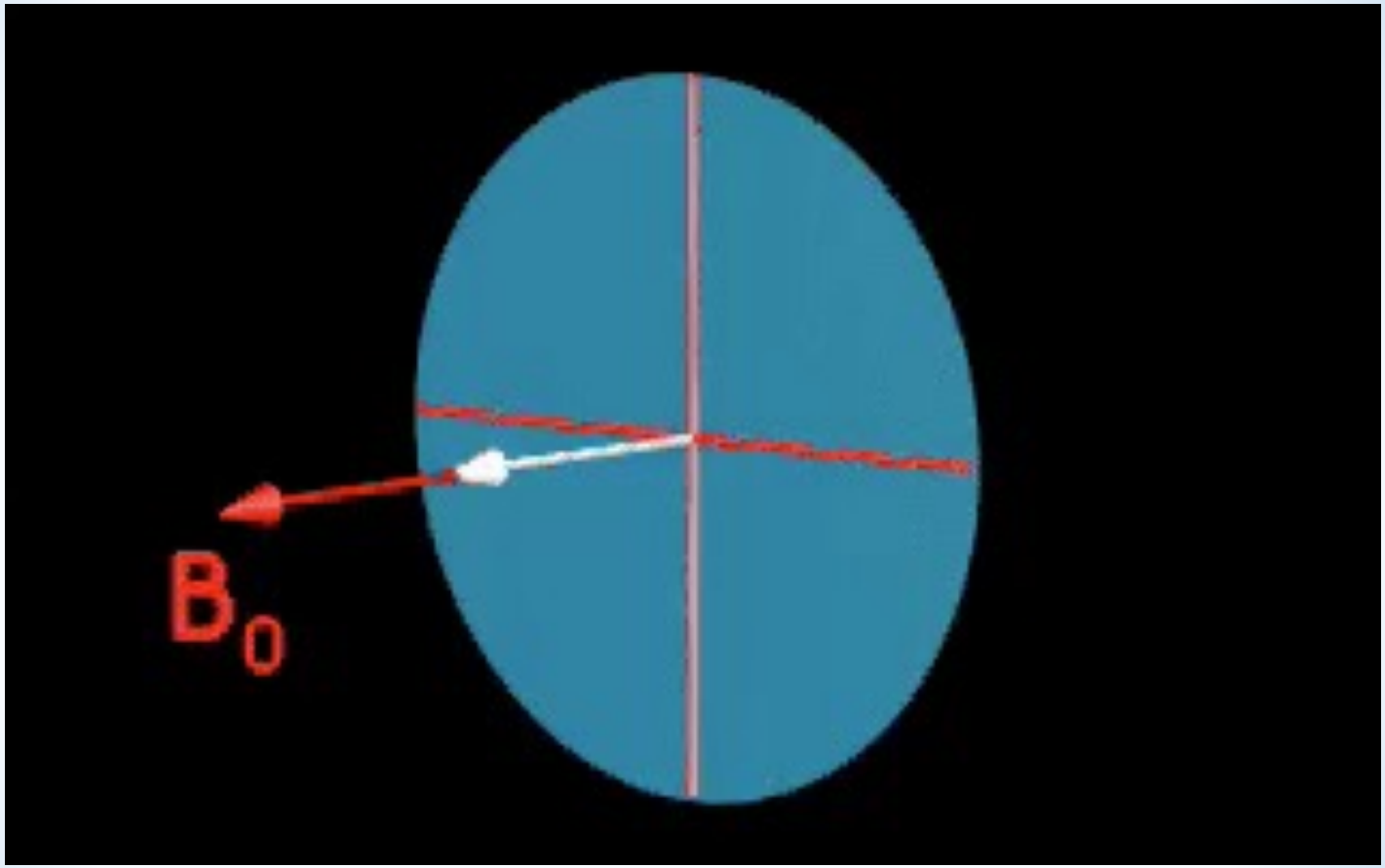
Rephase of spins after 180° RF- pulse in “rotating frame”



Effect of 180-pulse



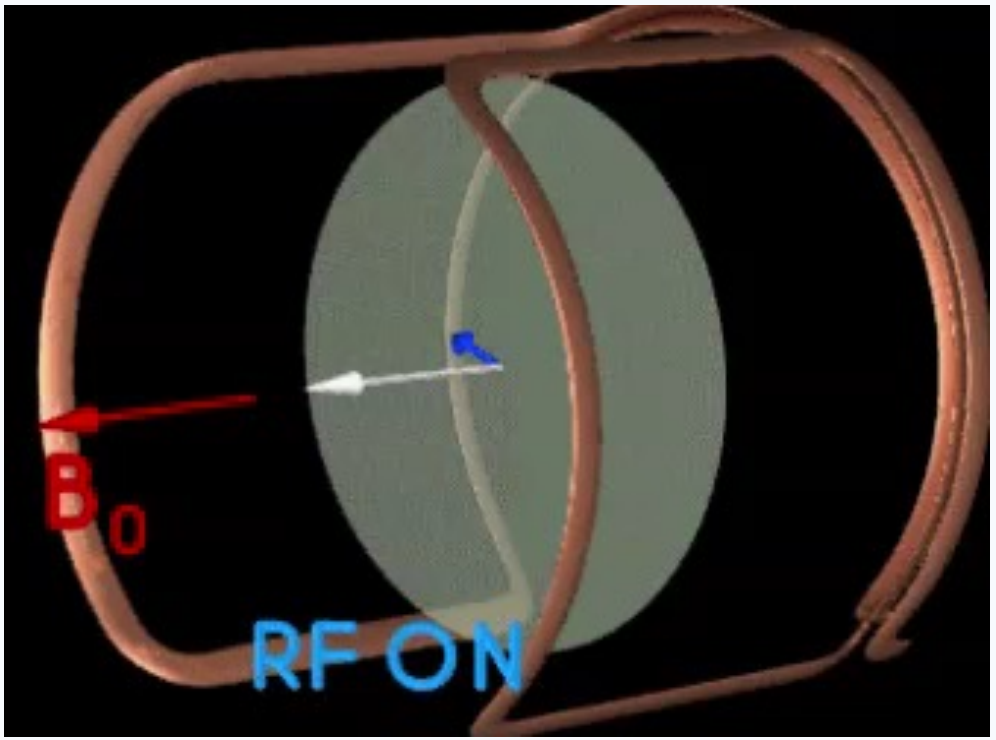
Spin Echo (180 degree pulse)



Relaxation

What happens to Magnetization after excitation?

Relaxation



T1-relaxation: recovery of longitudinal magnetization
T2-relaxation: decay of transversal magnetization

Transversal and longitudinal relaxation times

Relaxation is the dynamic physical process in which the system of spins returns to equilibrium. Relaxation can be broken down into:

- Recovery of longitudinal magnetization (M_z) aligned with B_0, following an exponential curve characterized by time constant T_1

  (Longitudinal (spin-lattice) **relaxation time T_1)**

- Decay of transverse magnetization (M_{xy}), due to spins getting out of phase, according to an exponential curve characterized by time constant T_2.

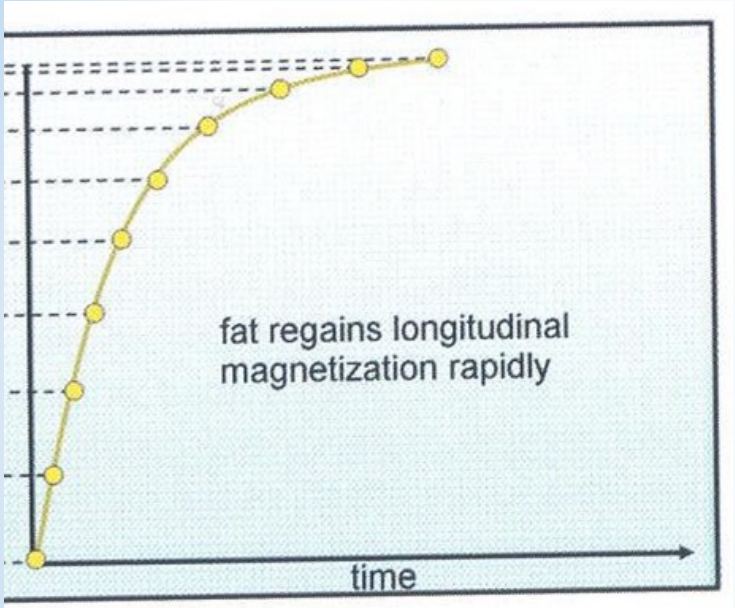
  (Transversal (spin-spin) **relaxation time T_2)**

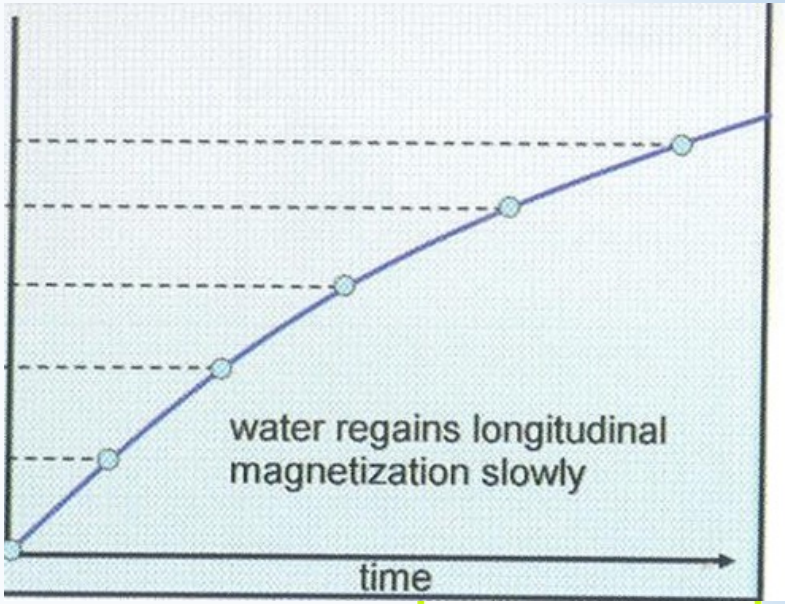
What is causing T1 relaxation?

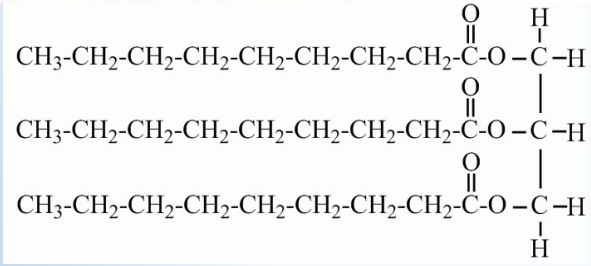
Variety of relaxation mechanisms allow nuclear spins to exchange energy with their surroundings, the lattice, allowing the spin populations to equilibrate. The fact that T1 relaxation involves an interaction with the surroundings is the origin of the alternative description, spin-lattice relaxation.

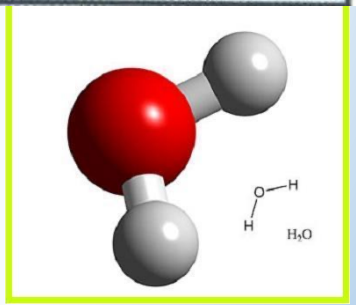
Note that the rates of T1 relaxation are generally strongly dependent on the NMR frequency and so vary considerably with magnetic field strength B. Small amounts of paramagnetic substances in a sample speed up relaxation very much.

T1 relaxation
(fat vs water)



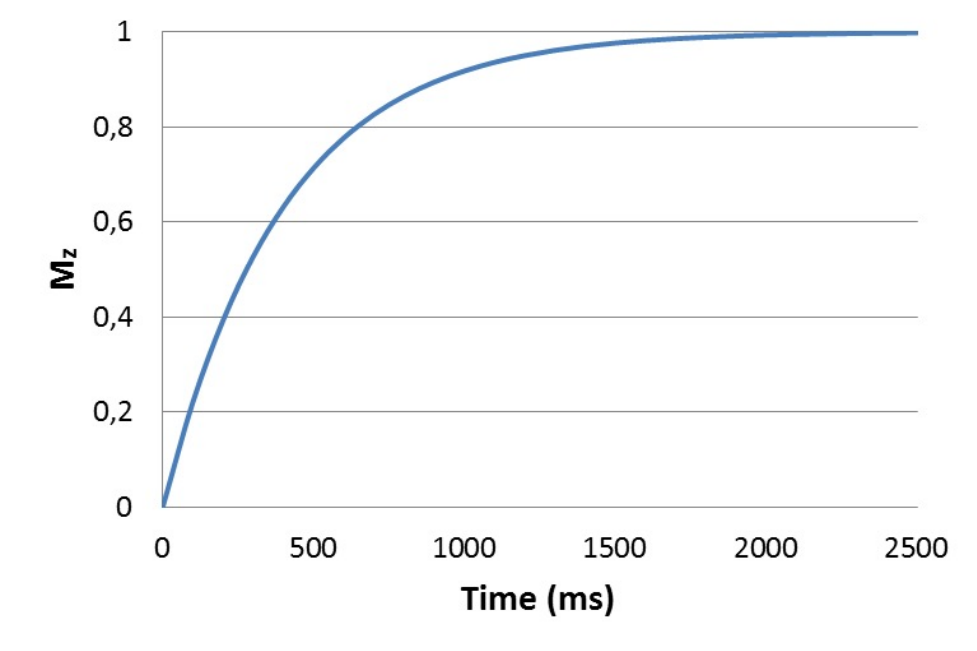






T1 relaxation time

Tissue T_1-relaxation time 400 ms



TR (ms)	M_z
200	39,3%
400	63,2%
800	86,5%
1200	95,0%
2000	99,3%

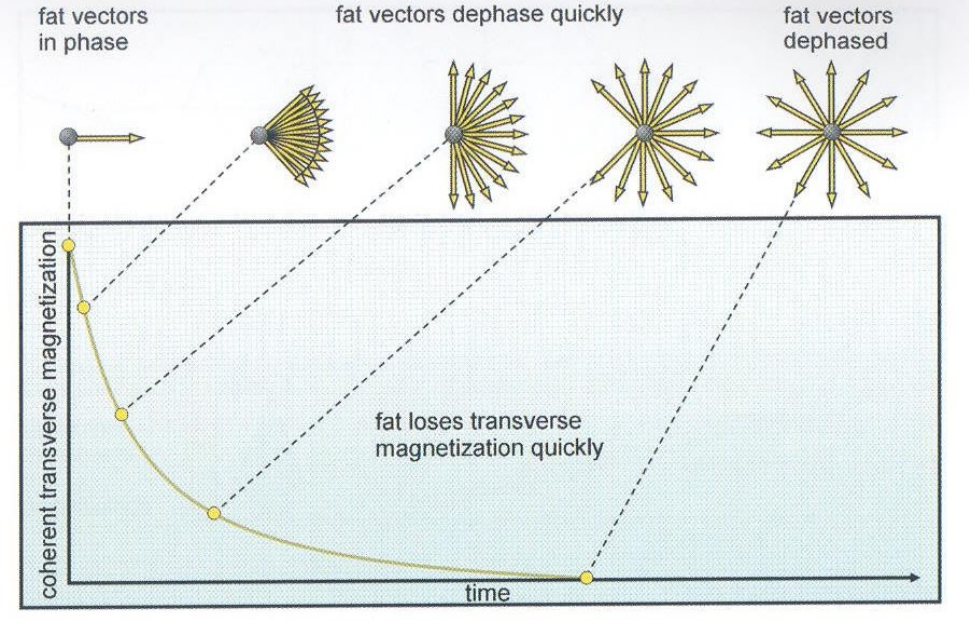
T1 relaxation: $M_z(t) = M_0 [1-\exp(-t/\mathbf{T1})]$

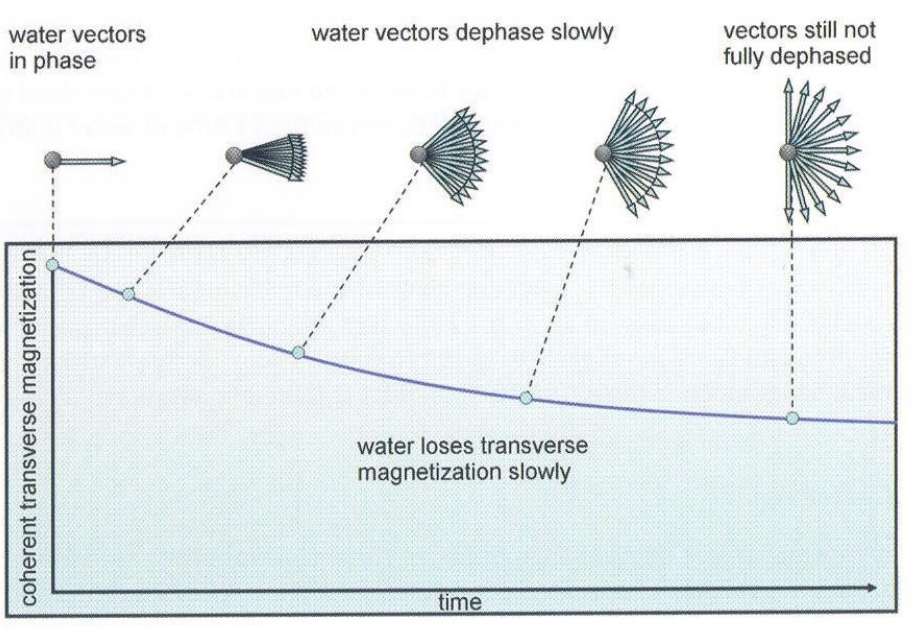
What is causing T2 relaxation?

T2 relaxation is a complex phenomenon, but at its most fundamental level, it corresponds to a decoherence of the transverse nuclear spin magnetization. Random fluctuations of the local magnetic field lead to random variations in the instantaneous NMR precession frequency of different spins. As a result, the initial phase coherence of the nuclear spins is lost, until eventually the phases are disordered and there is no net xy magnetization.

Because T2 relaxation involves only the phases of other nuclear spins it is often called "spin-spin" relaxation.

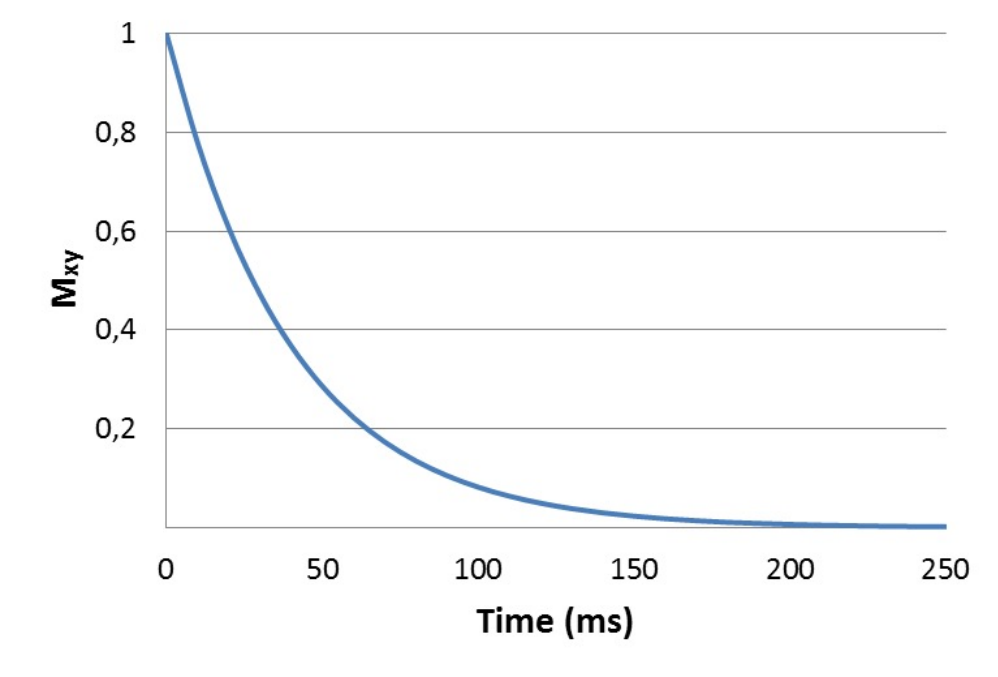
T2-relaxation
(fat vs water)





T2 relaxation time

Tissue T_2-relaxation time 40 ms

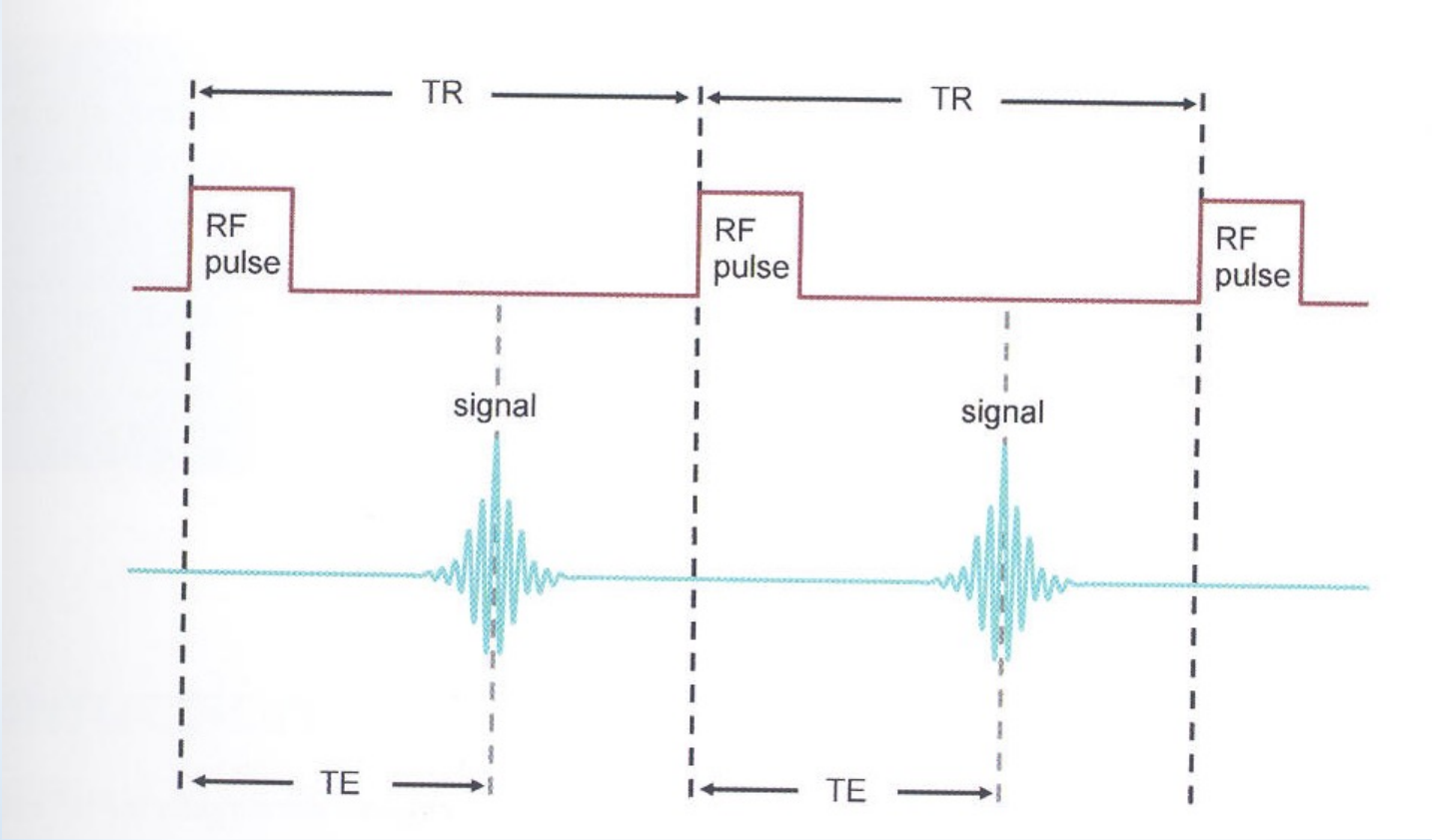


TE(ms)	M_{xy}
20	60.7%
40	36,8%
80	13,5%
120	5,0%
200	0,7%

T2 relaxation: $M_{xy}(t) = M_0 \exp(-t/\mathbf{T2})$

Contrast

Pulse sequence (simplest case)



Contrast

Pulse sequence – Spin Echo sequence (simple case)

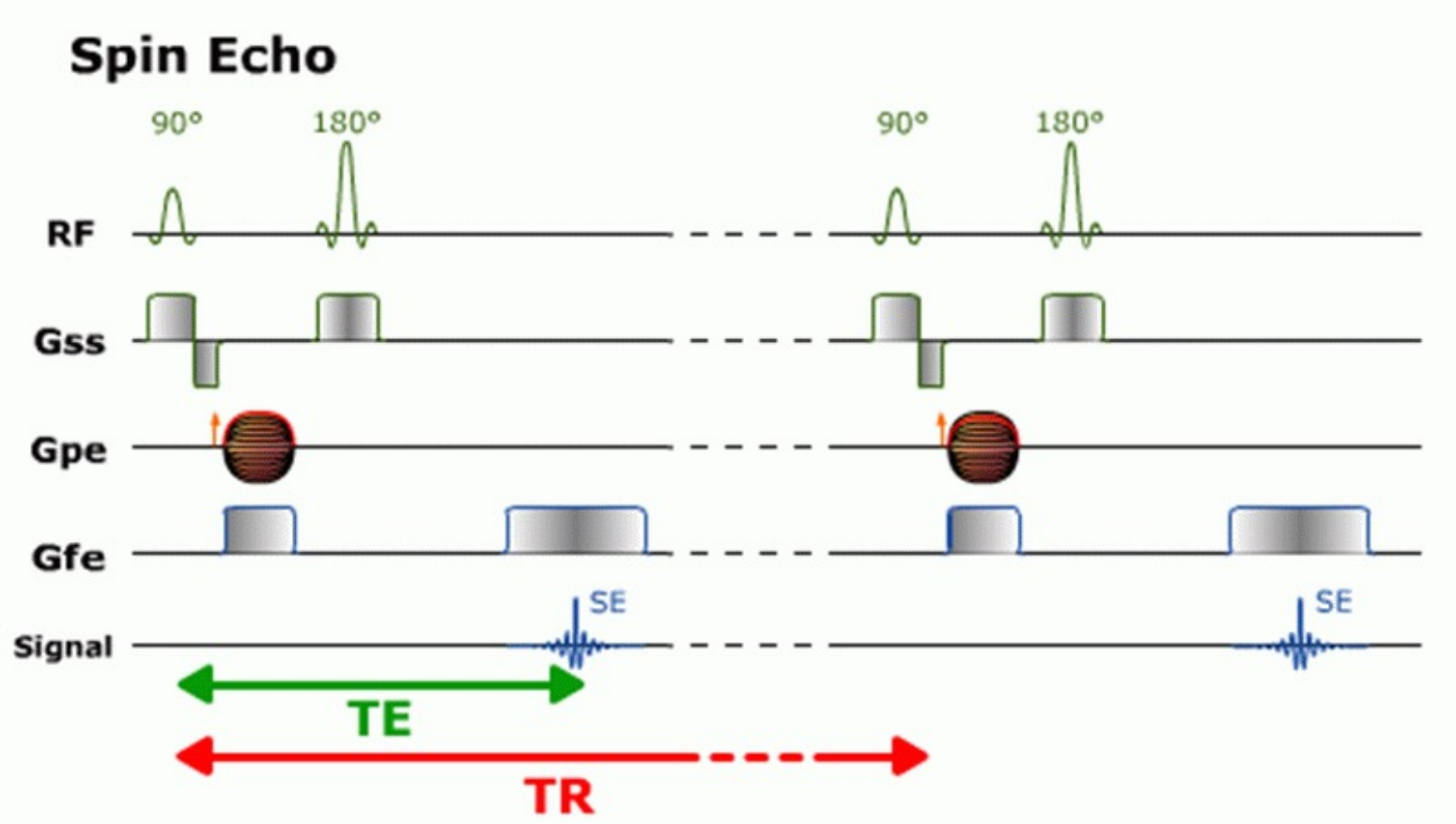
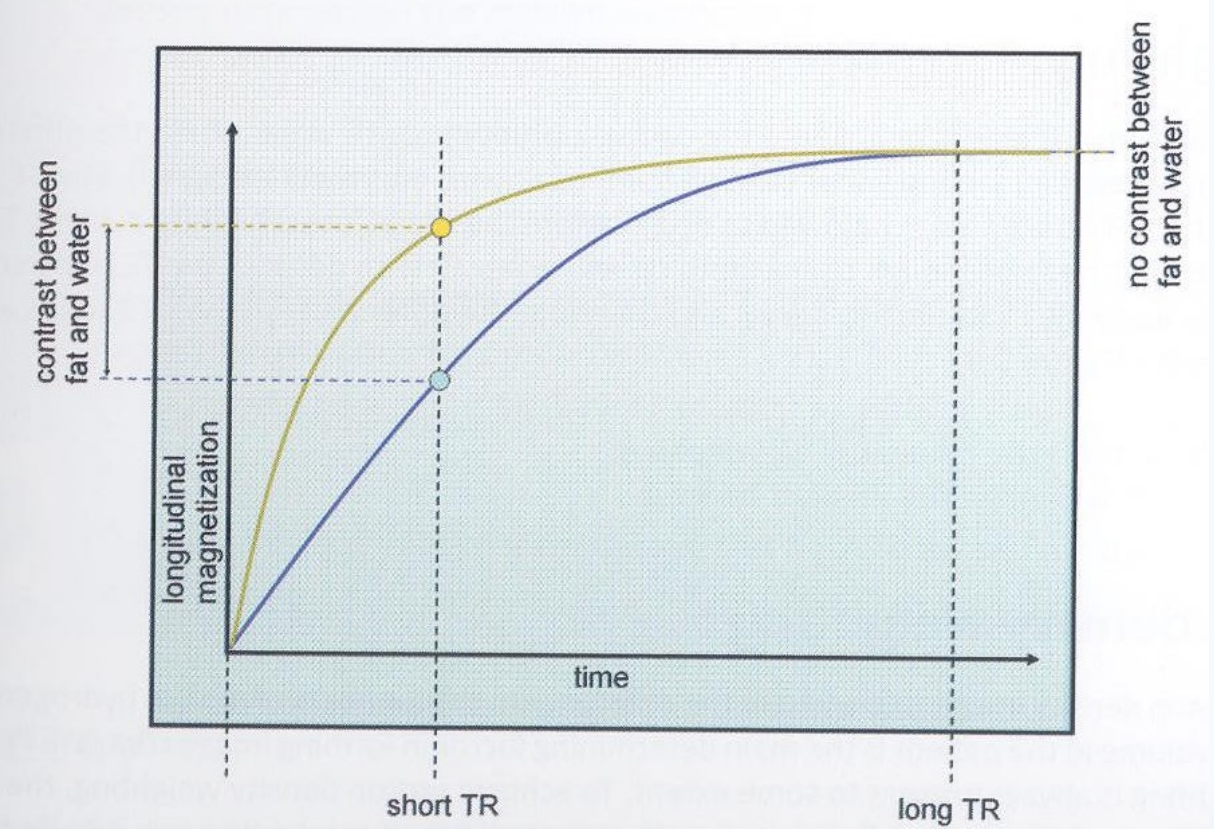
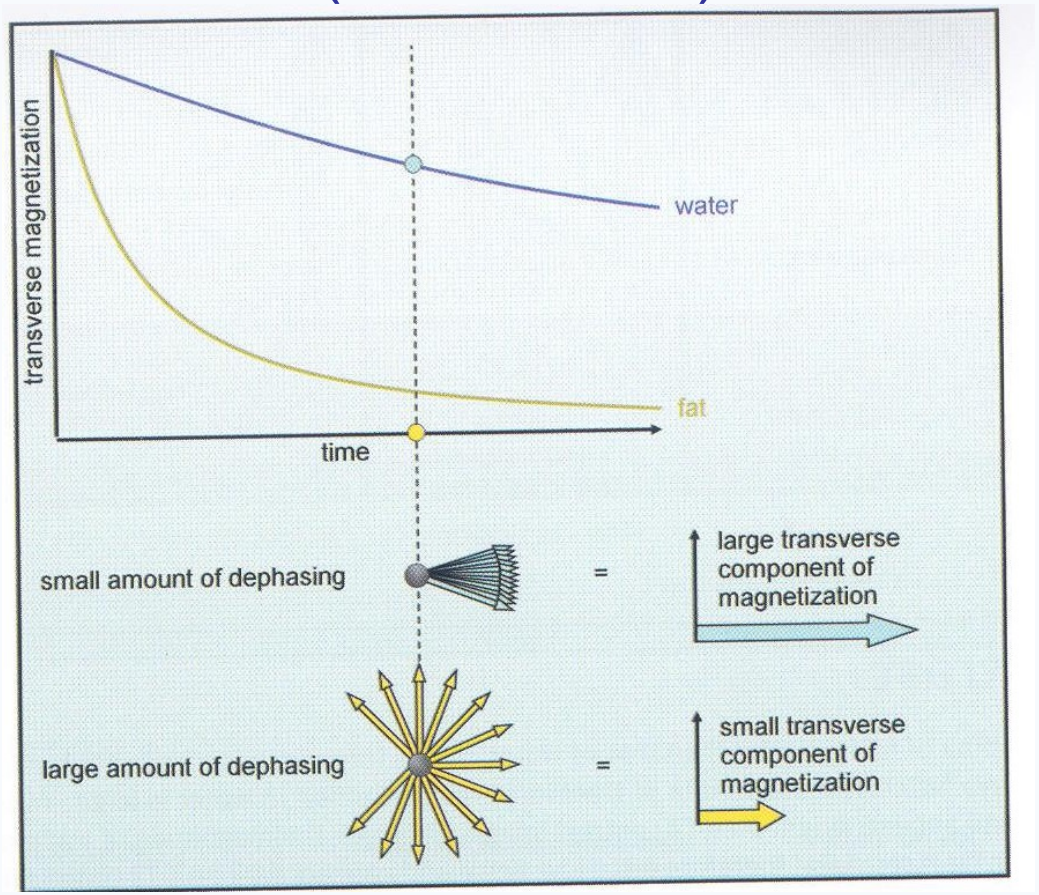


Image contrast and contrast agents

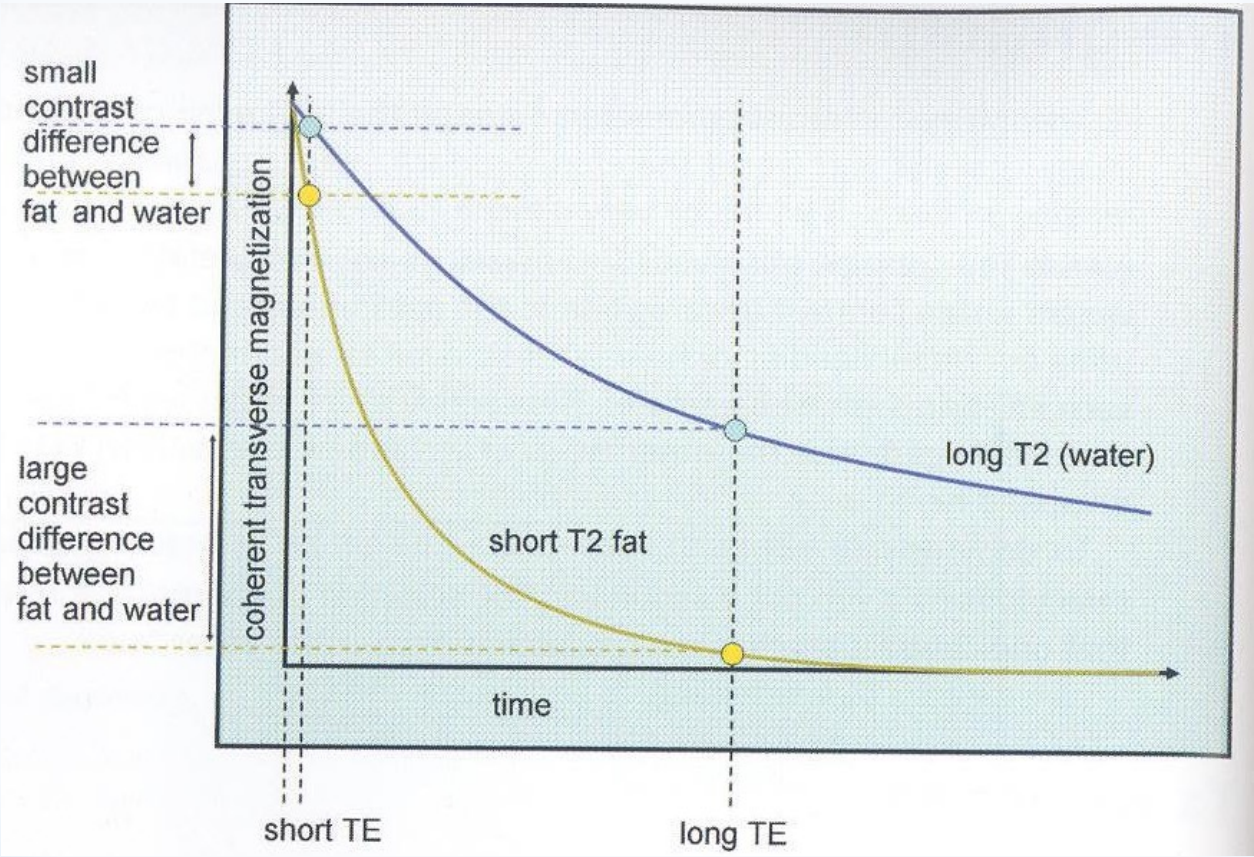
T1-contrast
(fat vs water)



T2-contrast
(fat vs water)



T2-contrast
(fat vs water)



Relaxation times in tissue

Table 2.1 T1 and T2 relaxation times of brain tissue at 1 T.

Tissue	T1 time (ms)	T2 time (ms)
Water	2500	2500
Fat	200	100
Cerebrospinal fluid	2000	300
White matter	500	100

- T1 weighted images are characterized by bright fat and dark water
- T2 weighted images are characterized by bright water and dark fat
- Proton density weighted images are characterized by areas with high proton density (bright) and areas with low proton density (dark)
- The T1 and T2 relaxation times of a tissue, although inherent to that tissue, are dependent on the field strength of the magnet. As field strength increases, tissues take longer to relax. Table 2.1 shows the T1 and T2 relaxation times of brain tissue at 1T

Properties of Tissue
PD, T1 ja T2

NMR-signal is proportional to amount of protons in the tissue.

"More protons, more signal"

T1 relaxation: $M_z(t) = M_0 [1-\exp(-t/\mathbf{T1})]$

T2 relaxation: $M_{xy}(t) = M_0 \exp(-t/\mathbf{T2})$

T1 and T2 relaxation times are properties of a tissue

“Native” contrast in MR imaging

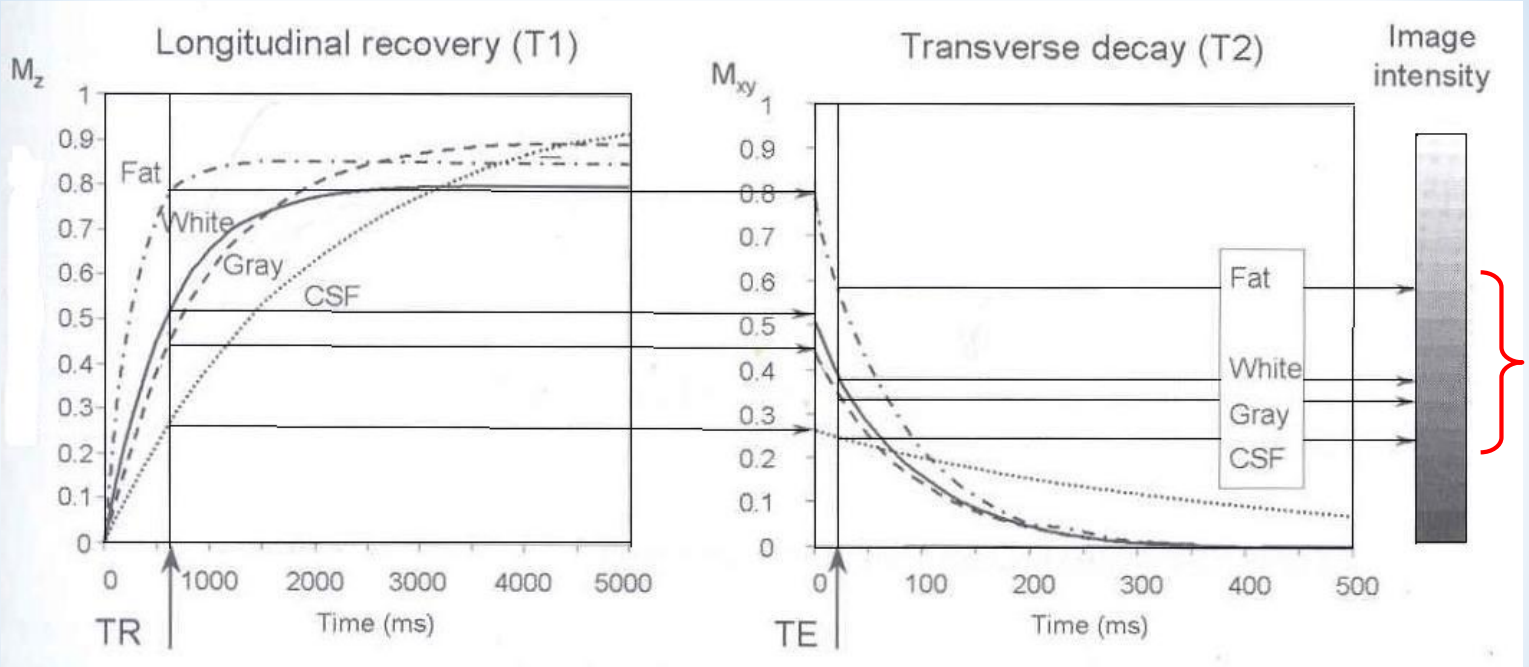
Tissue dependent parameters:

- PD: proton density
- T1: longitudinal (spin-lattice) relaxation time
- T2: transversal (spin-spin) relaxation time

Operator controllable parameters:

- TR : repetition time
- TE: echo time
- TI: inversion time
- FA: flip angle of M_Z

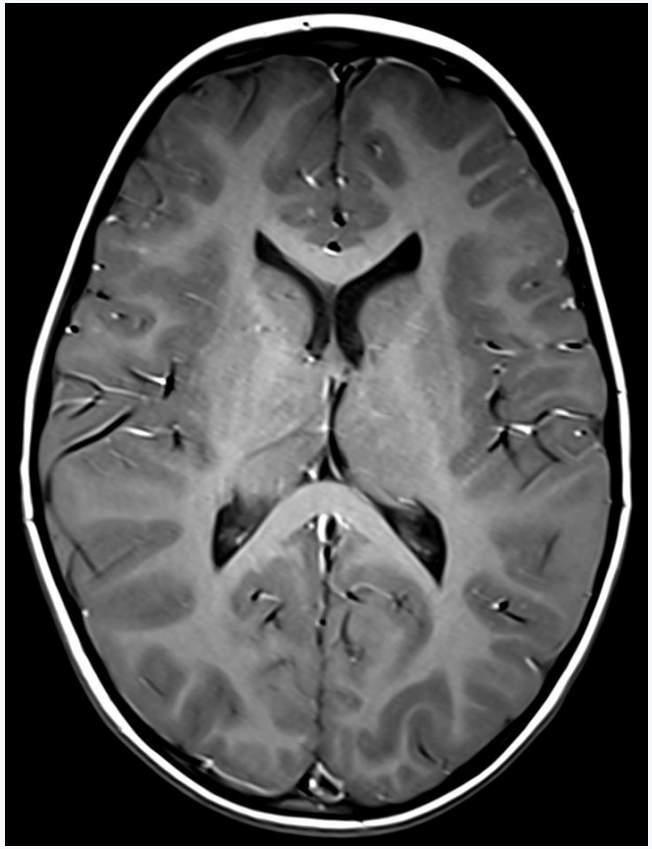
T1-weighted scan



Short TR (300 – 700 ms)

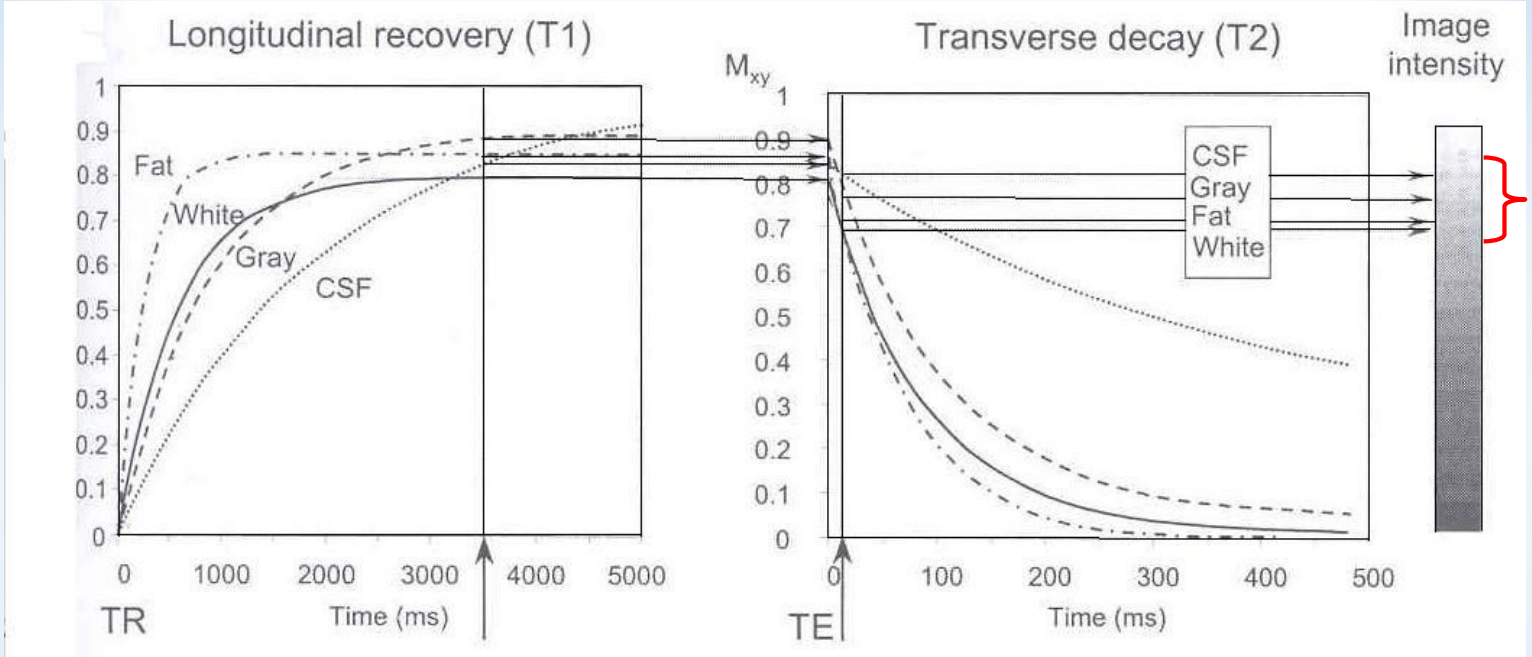
Short TE (5 – 25 ms)

Fat is ”bright” (high signal), CSF ”dark” (low signal)



T1-weighted image
TR = 600 ms, TE = 6,4 ms

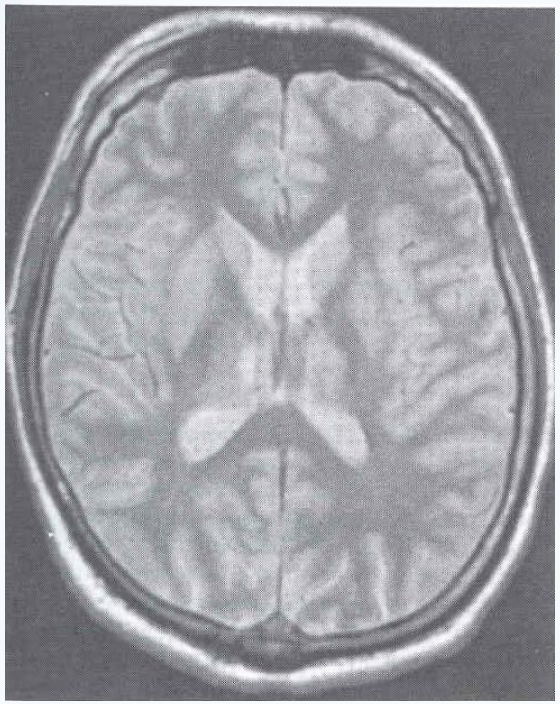
Proton density (PD) weighted scan



Long TR (2 – 4 s)

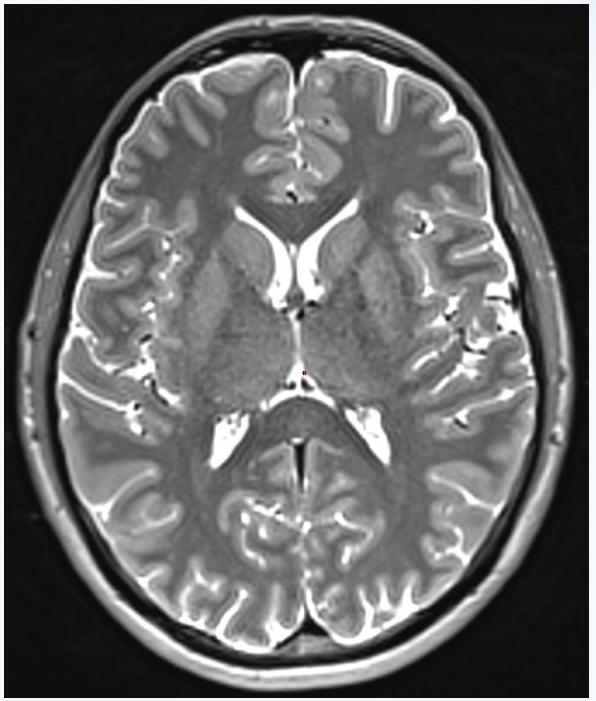
Short TE (5 – 30 ms)

Highest signal, but often weak contrast



PD-weighted

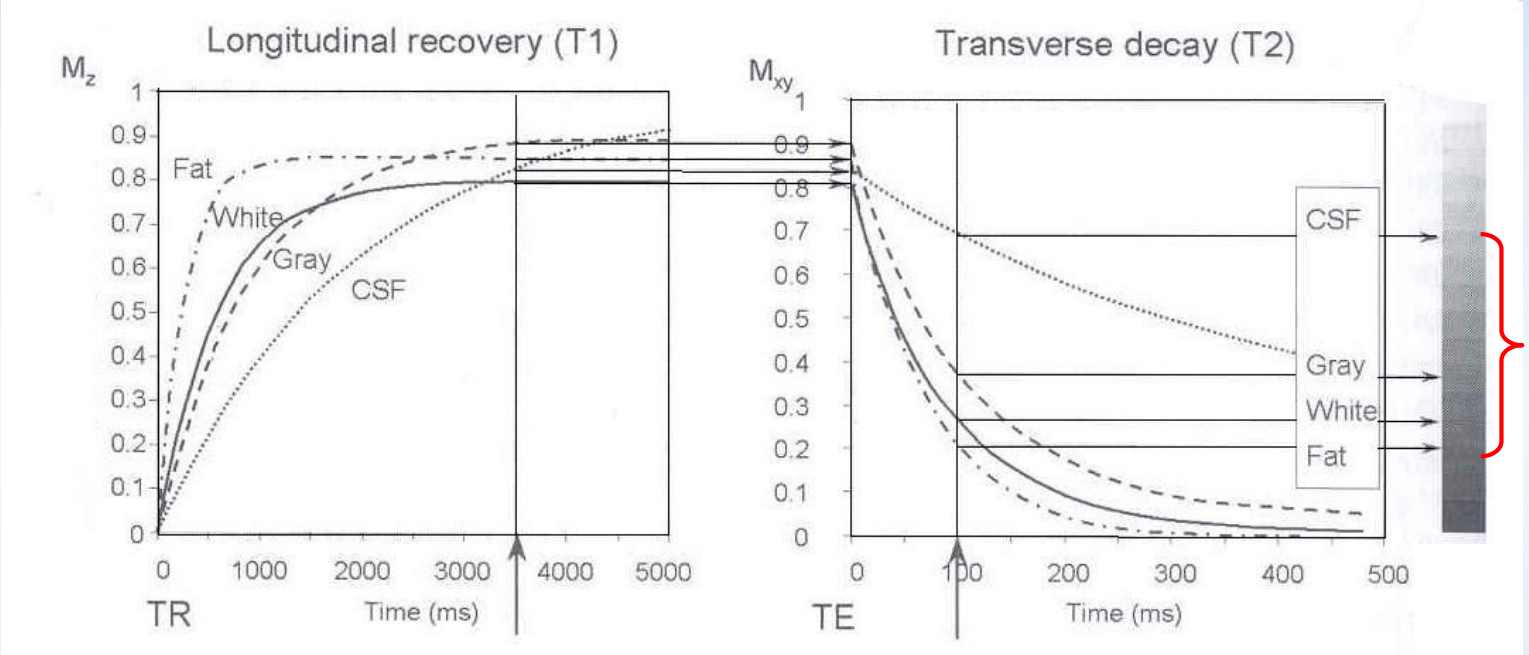
TR = 2400 ms, TE = 30 ms



PD-weighted TSE

TR = 12070 ms, TE = 13 ms, ETL 7

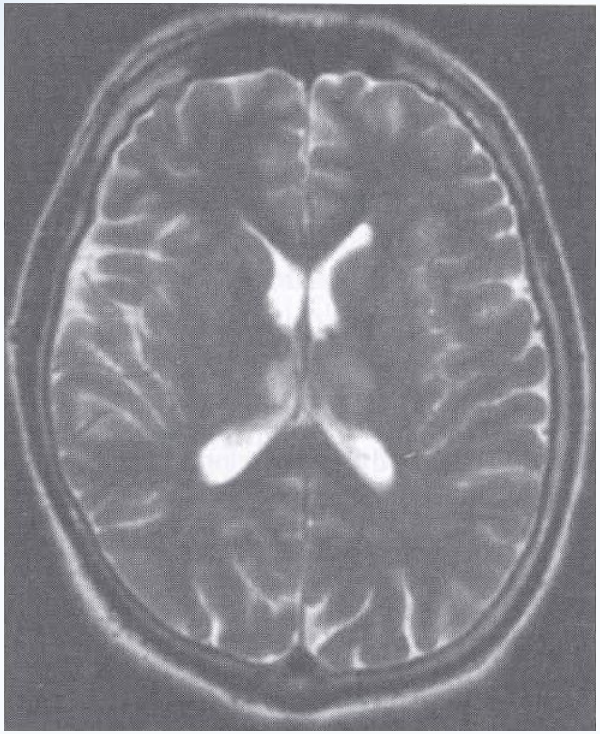
T2-weighted scan



Long TR (2 – 4 s)

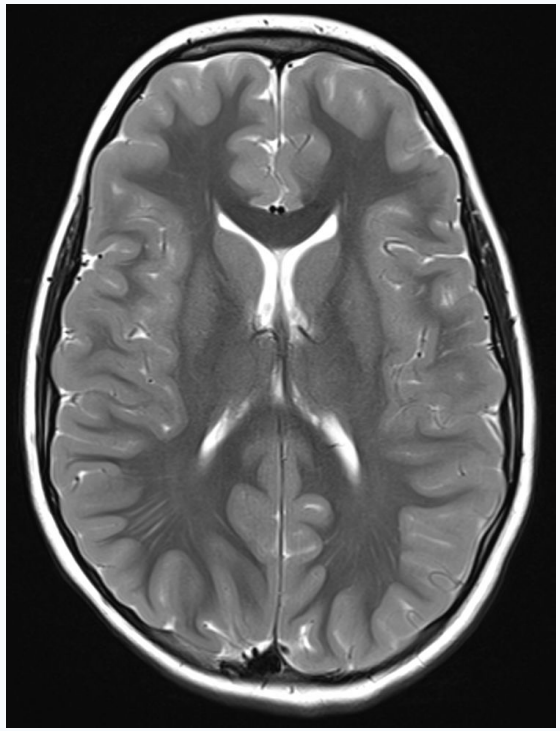
Long TE (60 – 120 ms)

CSF ”bright” (strong signal), fat ”dark” (weak signal)



T2-weighted SE

TR = 2400 ms, TE = 90 ms

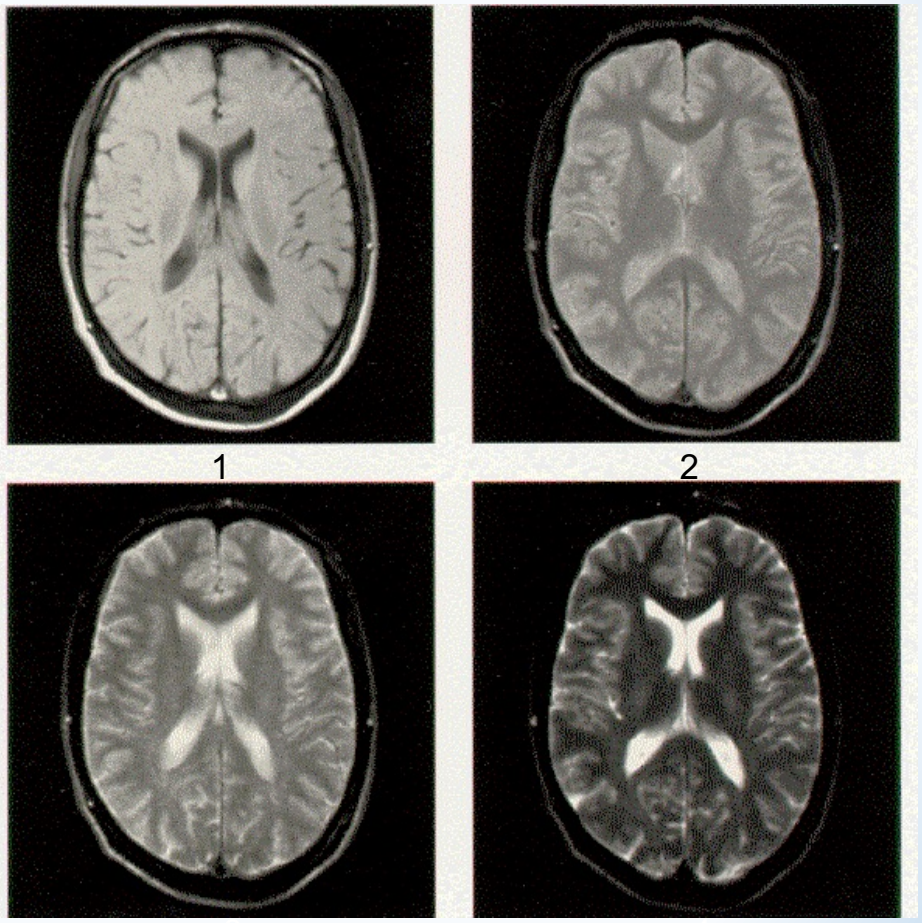


T2-weighted TSE

TR = 4120 ms, TE = 103 ms, ETL 11

T2 contrast in head MR images

	1	2	3	4
TR = 2 500				
TE =	15	60	90	120



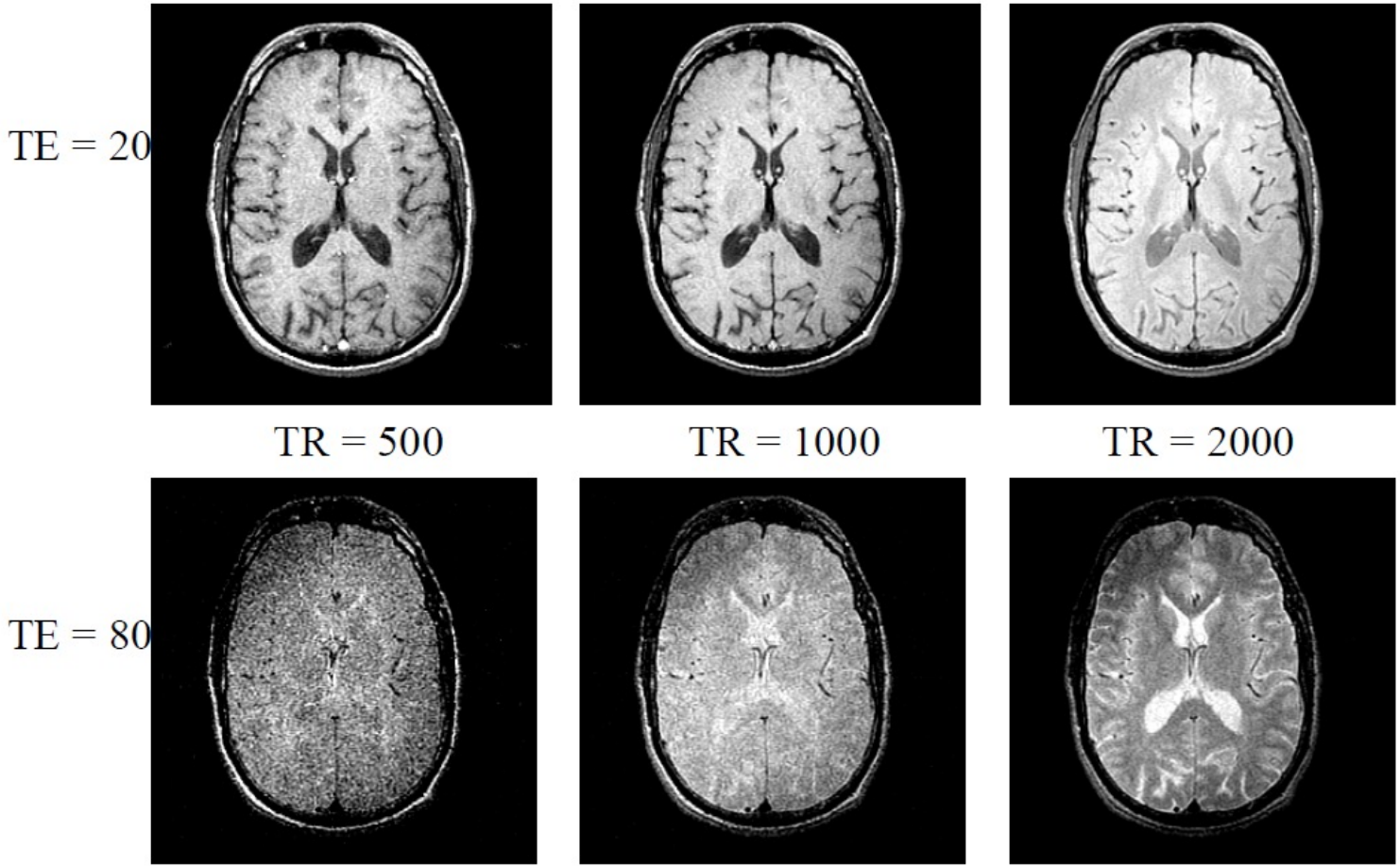
1: Proton density- weighted (PDW) image

2: Moderately T2- weighted (T2W) image

3: Routine T2- weighted (T2W) image

4: Heavily T2- weighted (T2W) image

Effect of TR and TE



Contrast agents

Paramagnetic contrast agents:

- Electron magnetic moment ~ 700*proton magnetic moment

- T1- relaxation effect very strong

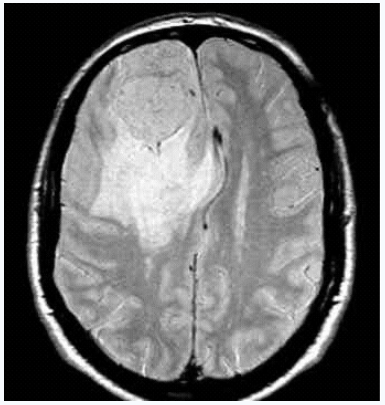
- Gd- DTPA: Gd^{3+} acts as a relaxing centre

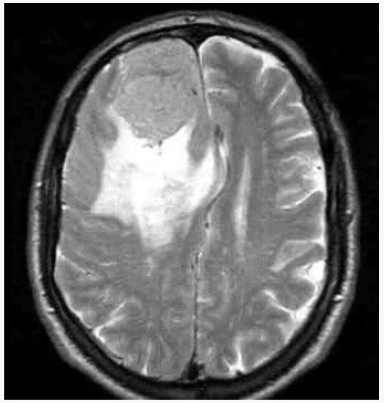
- T1 of enhanced lesion is shorter than T1 of normal tissue

- In T1W- MR image lesion seems **brighter** than normal tissue = **positive** contrast enhancement

MRI contrast in brain: an atypical meningioma

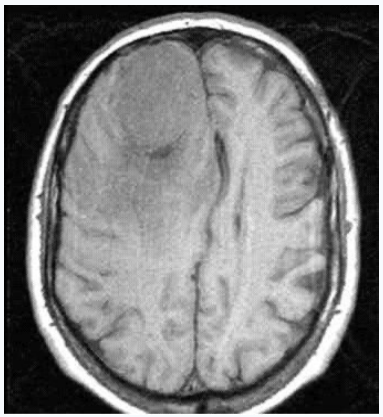
PD- weighted, PDW

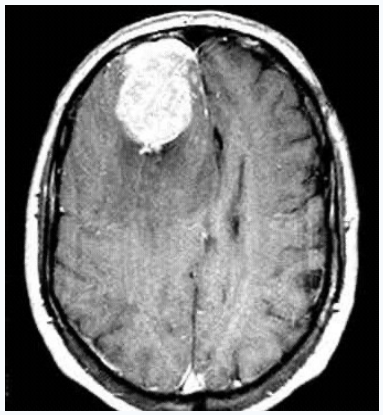




T2-weighted, T2W

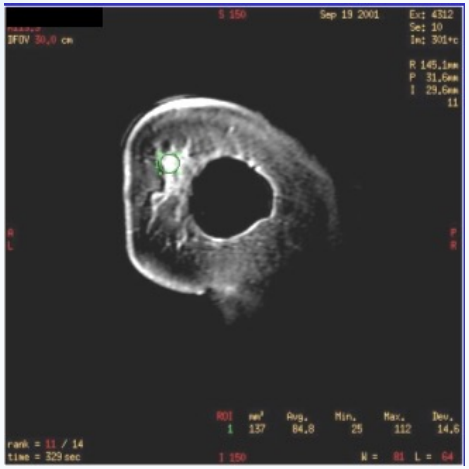
T1-weighted, T1W, no contrast agent



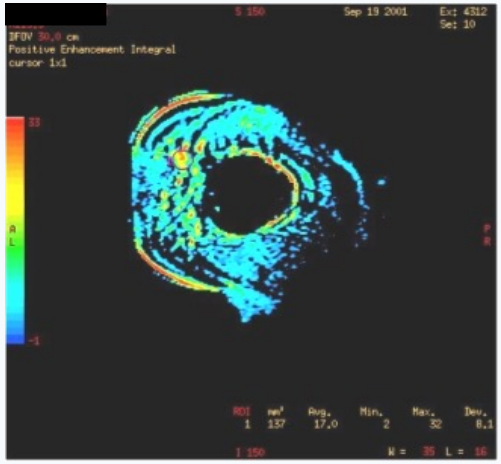


T1-weighted, T1W, With contrast agent

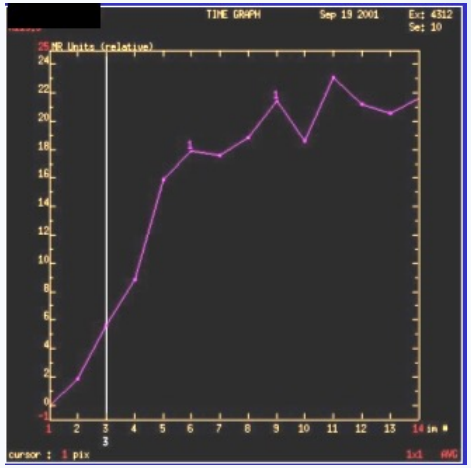
Dynamical CE Breast MRI



Gd-DTPA- enhanced magnitude image



Positive Enhancement Integral
PEI map

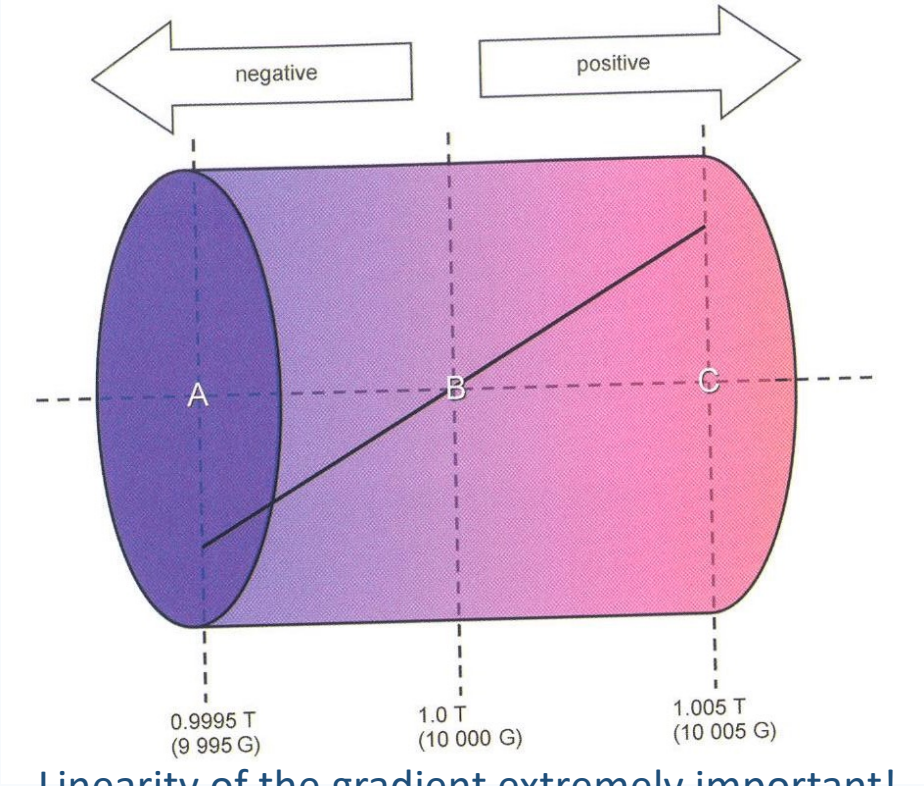


Intensity enhancement dynamical curve

From Signal to Image

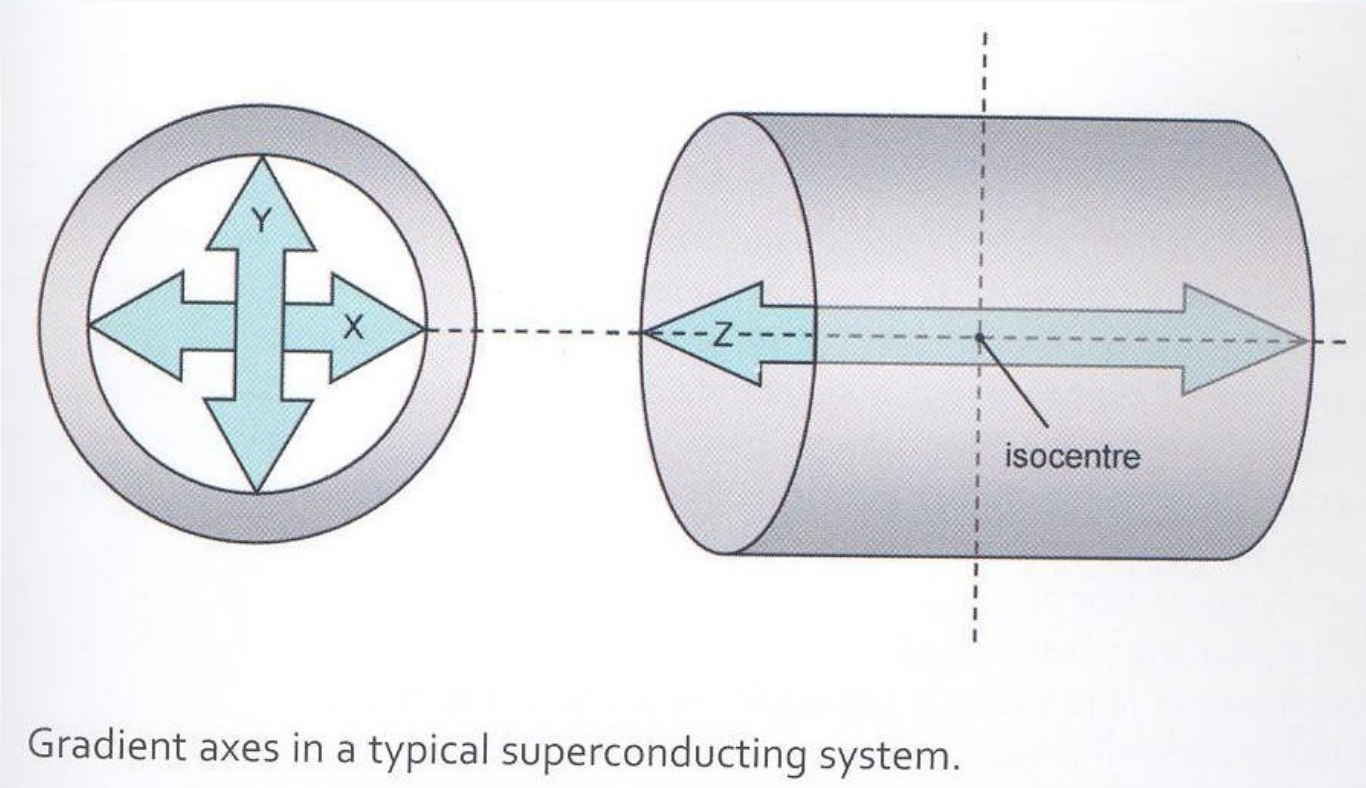
GRADIENTS

Magnetic field gradient

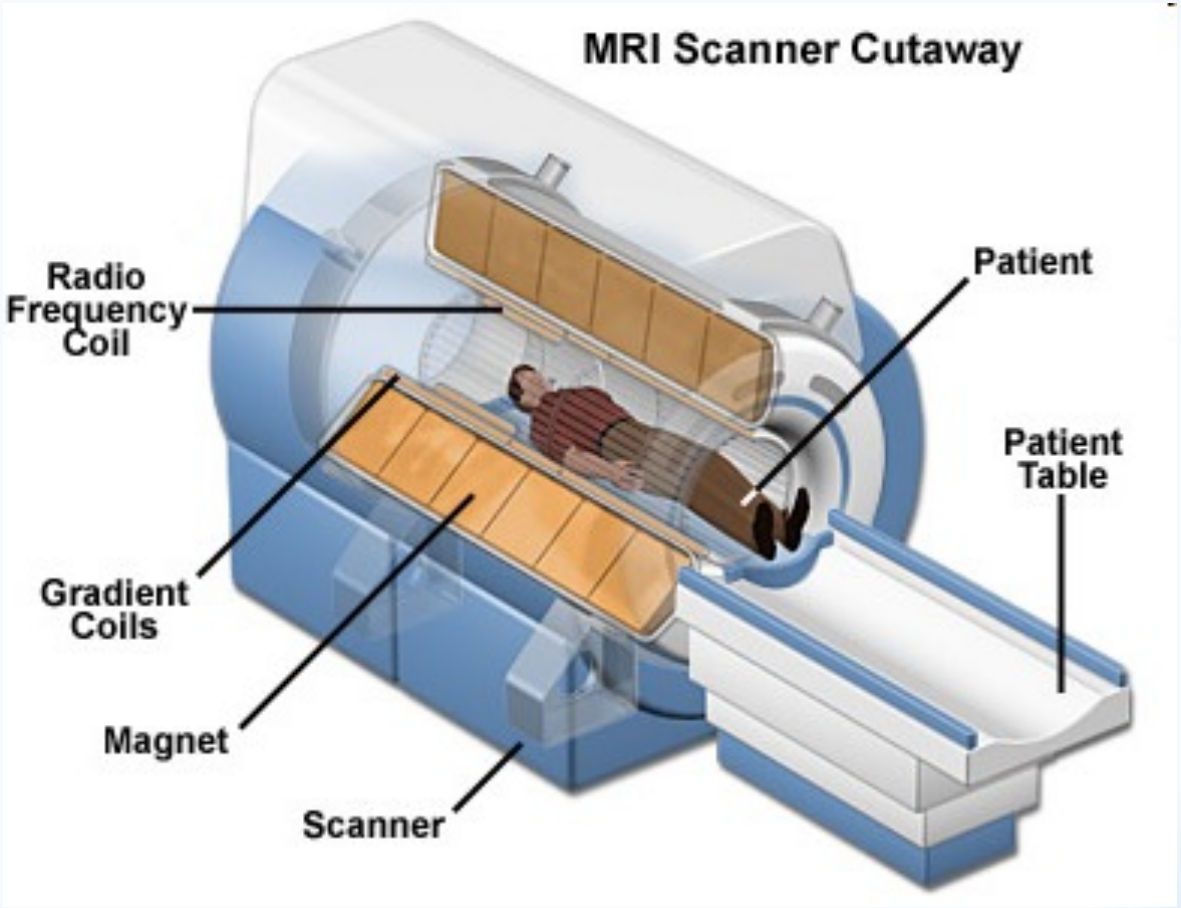


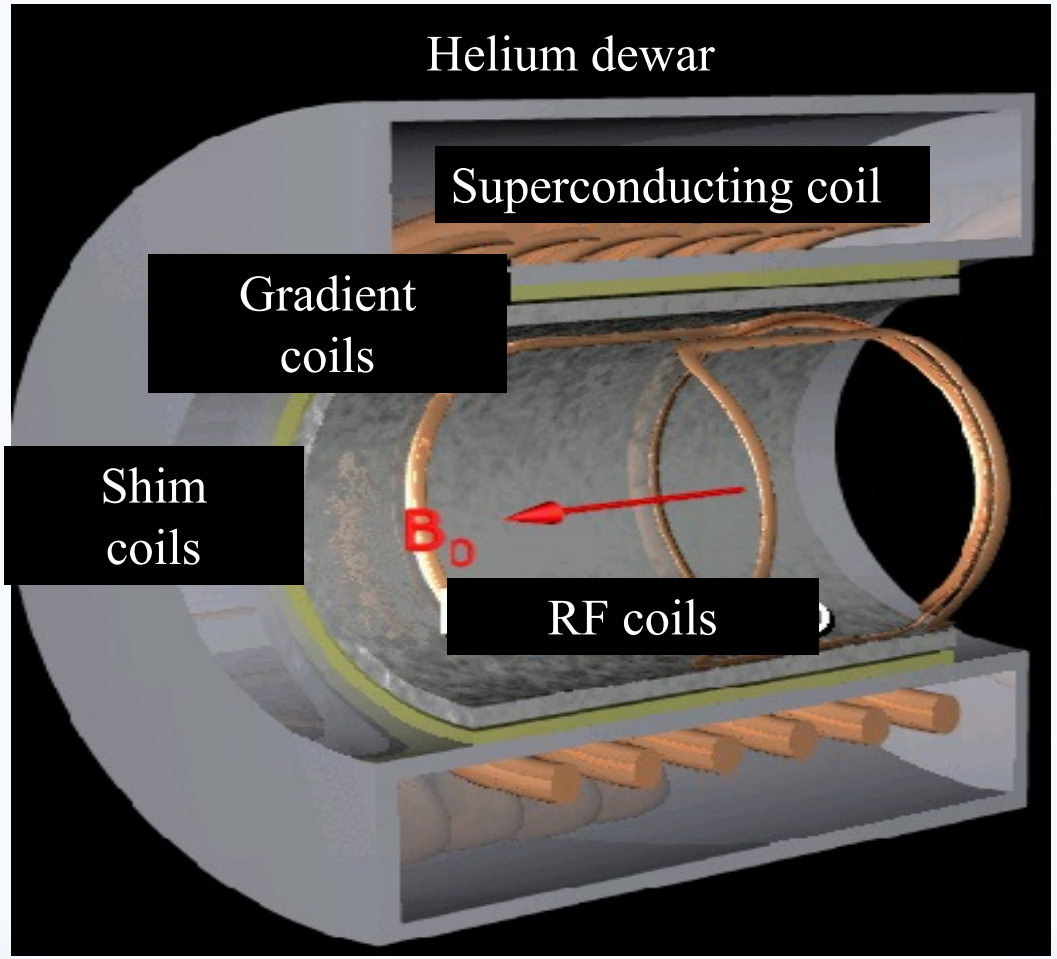
Linearity of the gradient extremely important!

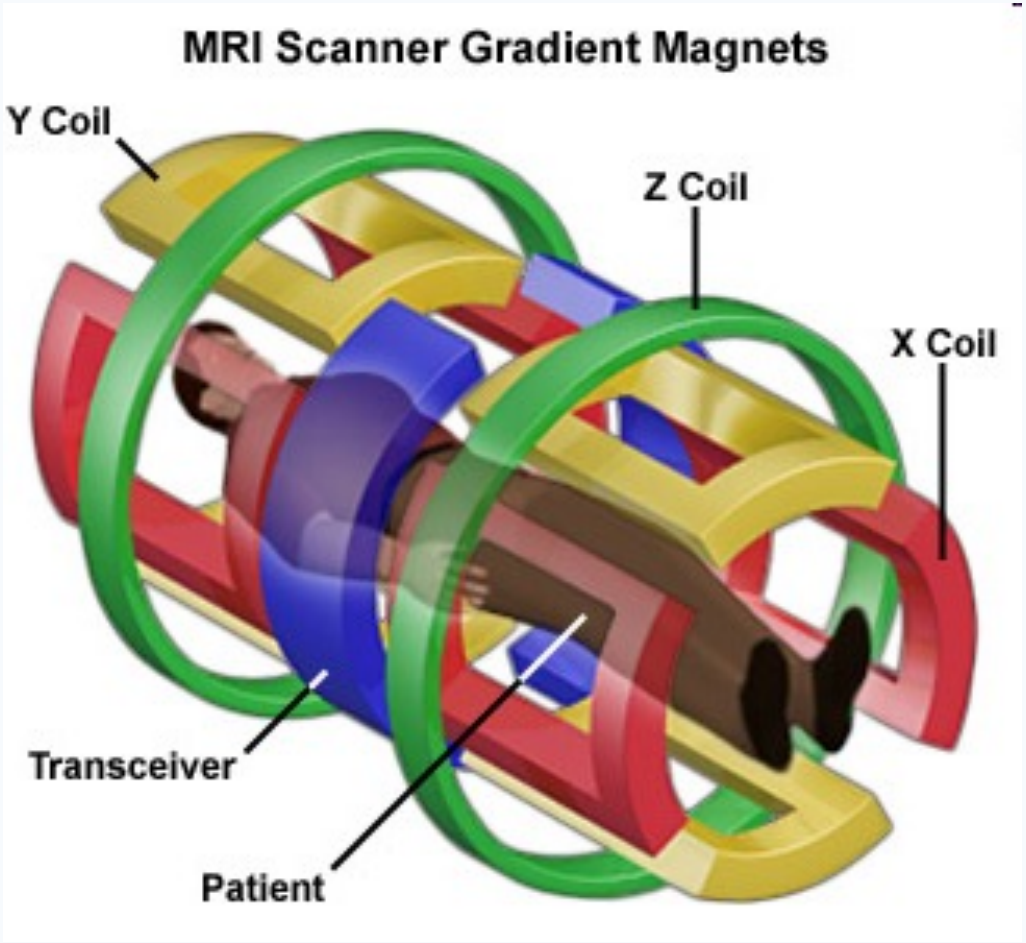
Gradient axis directions



Gradient axes in a typical superconducting system.

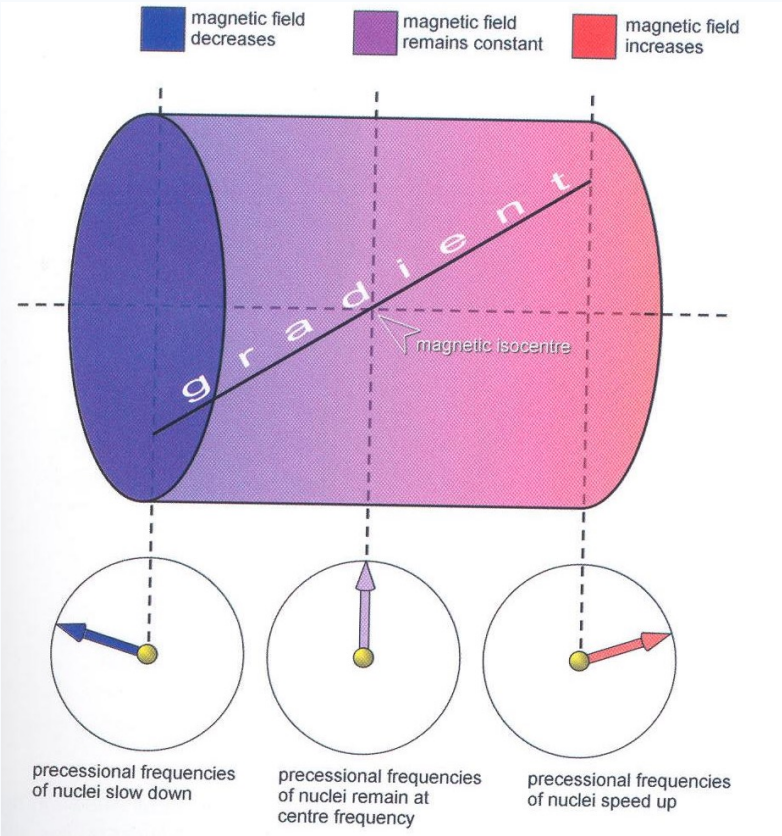

MRI Scanner Cutaway
Radio Frequency Coil
Patient
Patient Table
Gradient Coils
Magnet
Scanner


Helium dewar
Superconducting coil
Gradient coils
Shim coils
B_0
RF coils

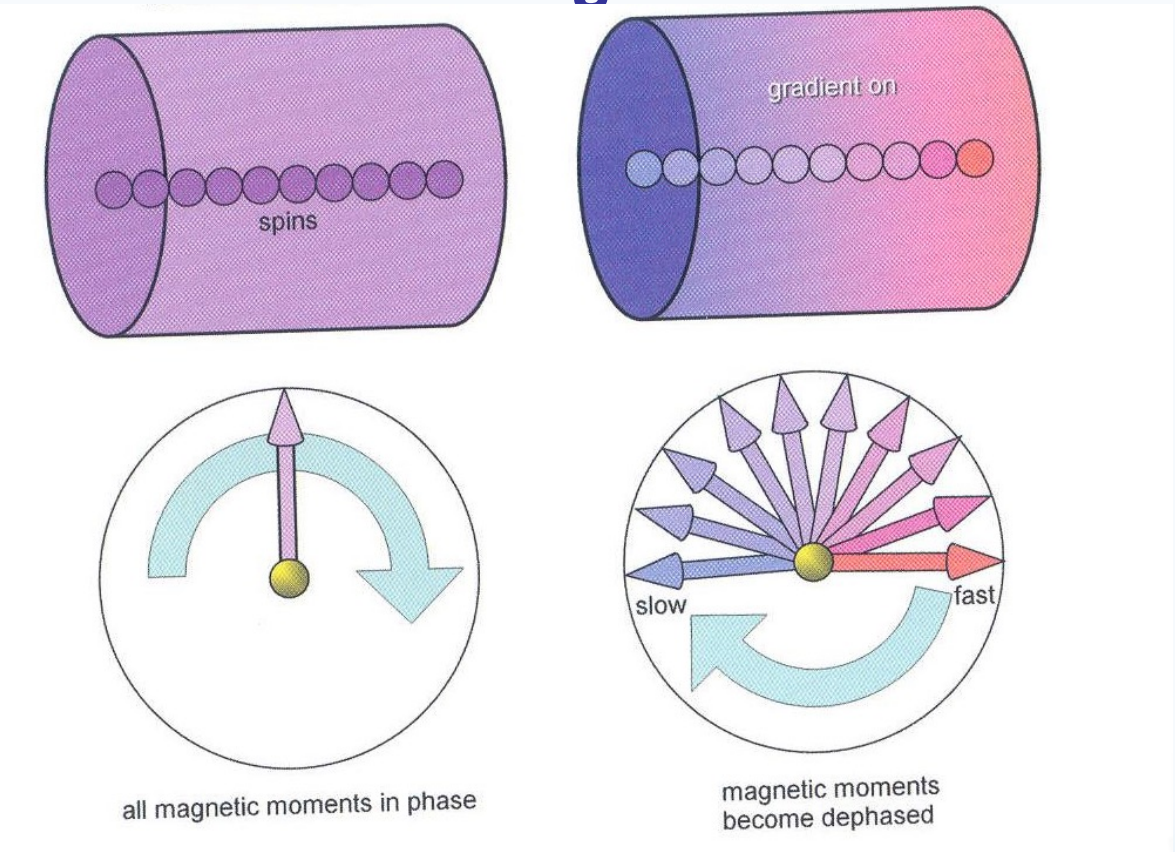

MRI Scanner Gradient Magnets
Y Coil
Z Coil
X Coil
Transceiver
Patient

Receiver coils

Magnetic field gradient

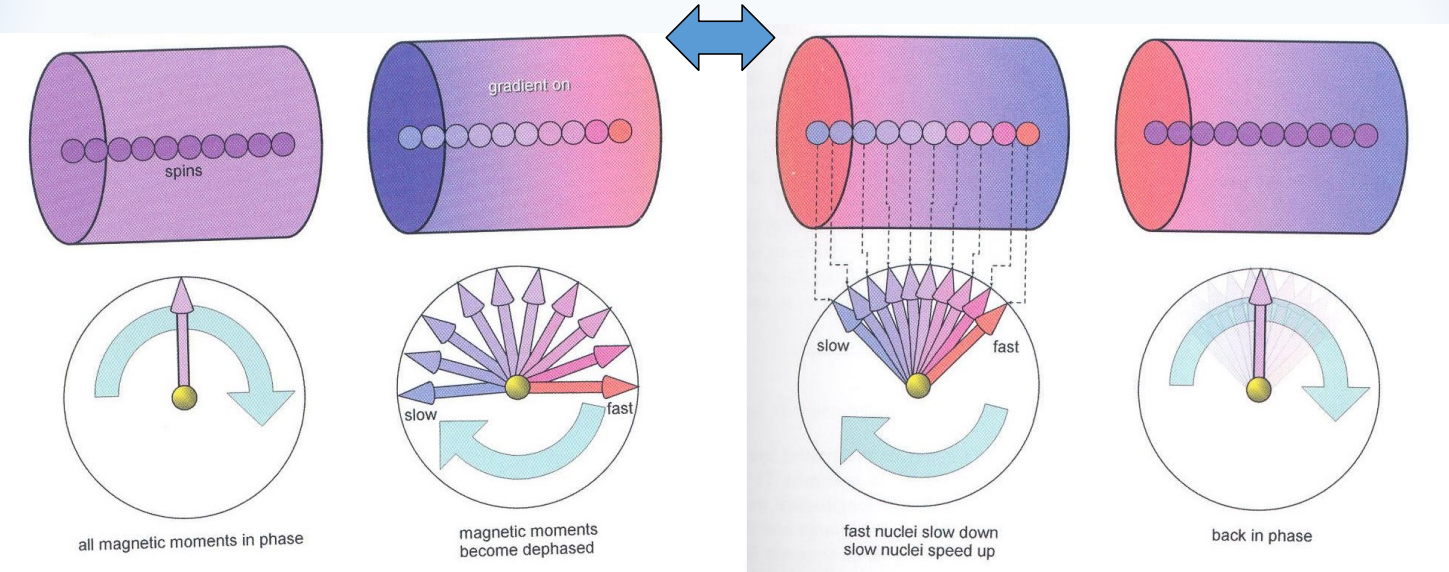


Effect of gradient



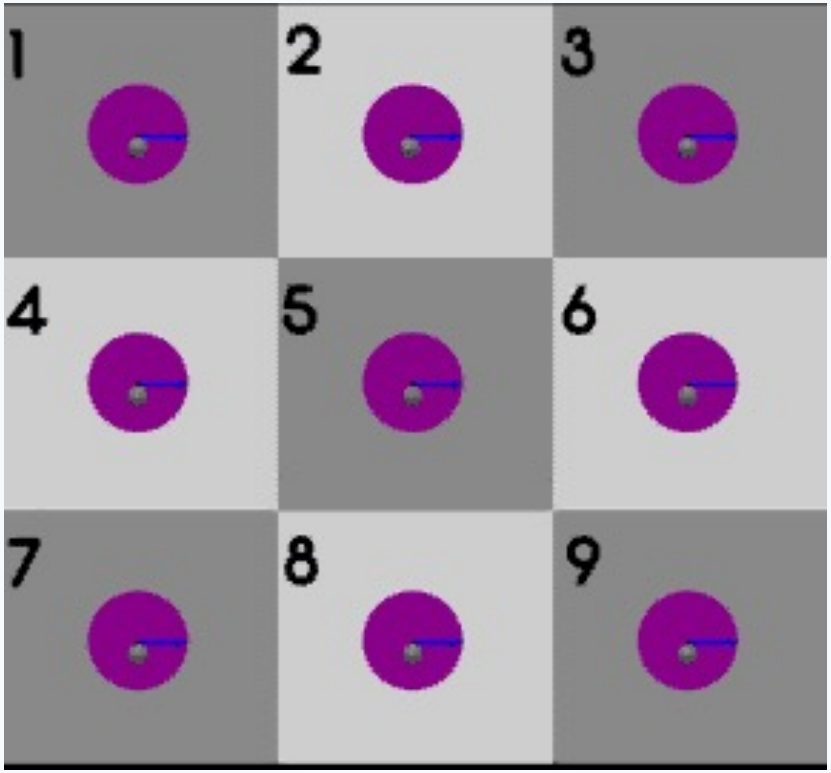
Gradient echo

Direction of the gradient switches

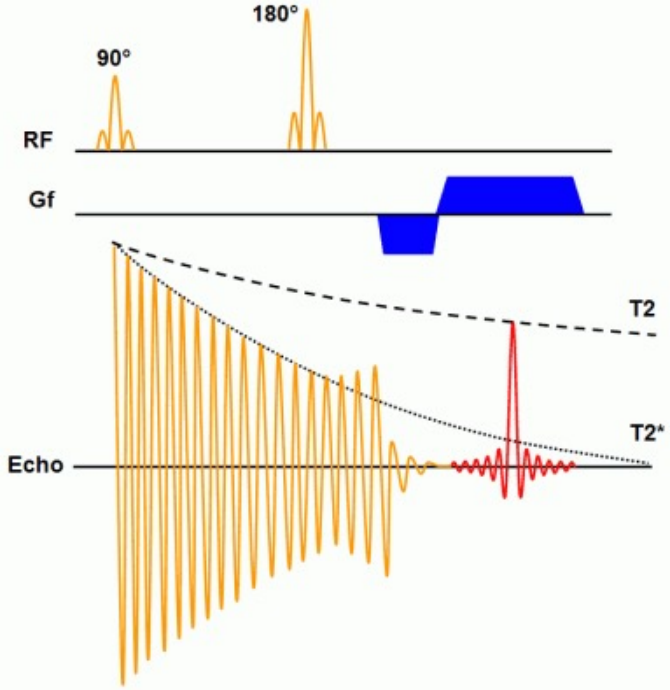


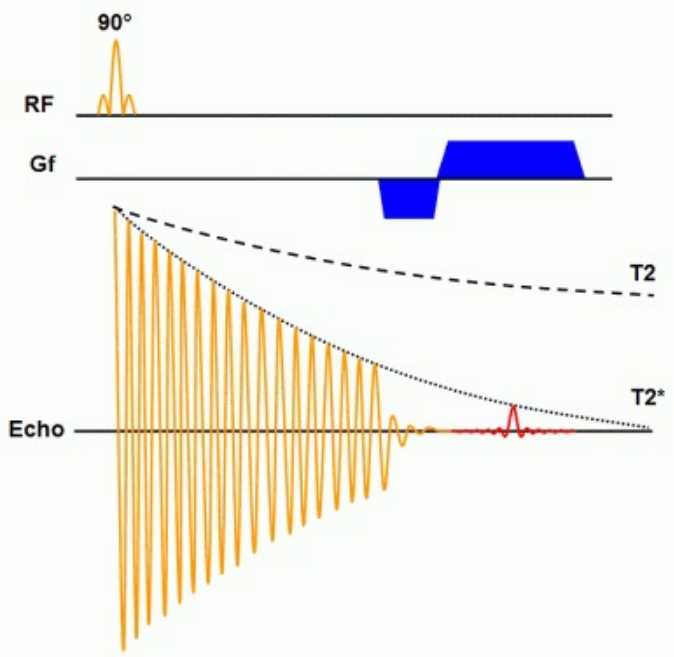
Gradient echo

Direction of the gradient switches



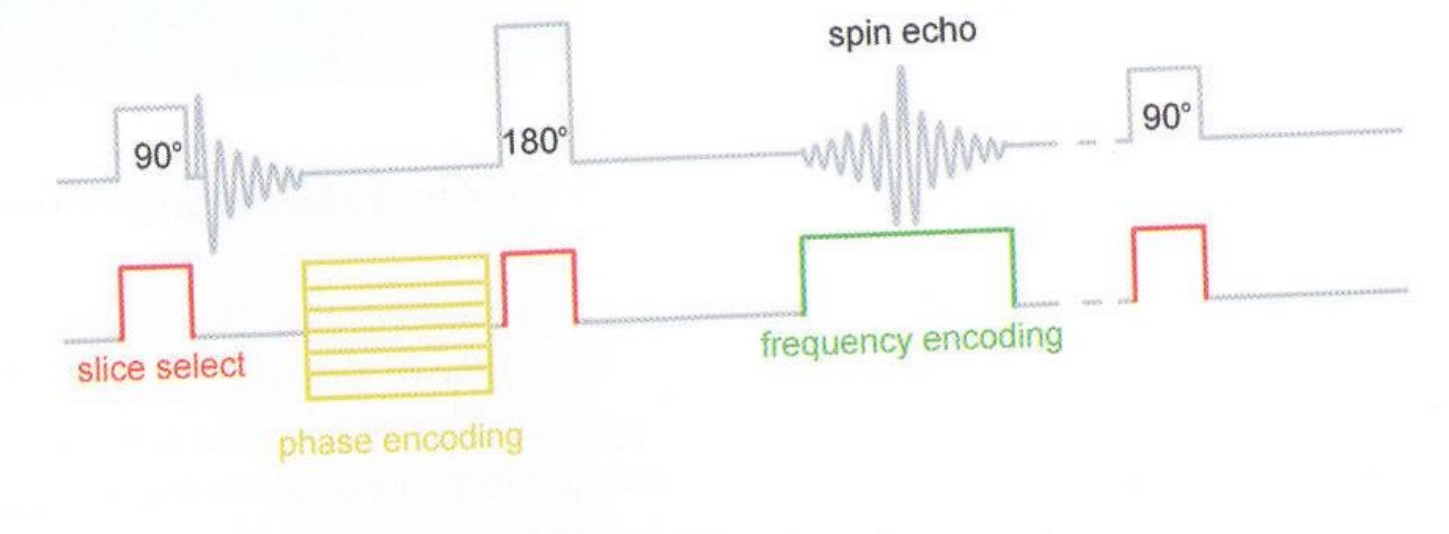
Signal from gradient echo is much weaker than one from spin echo:



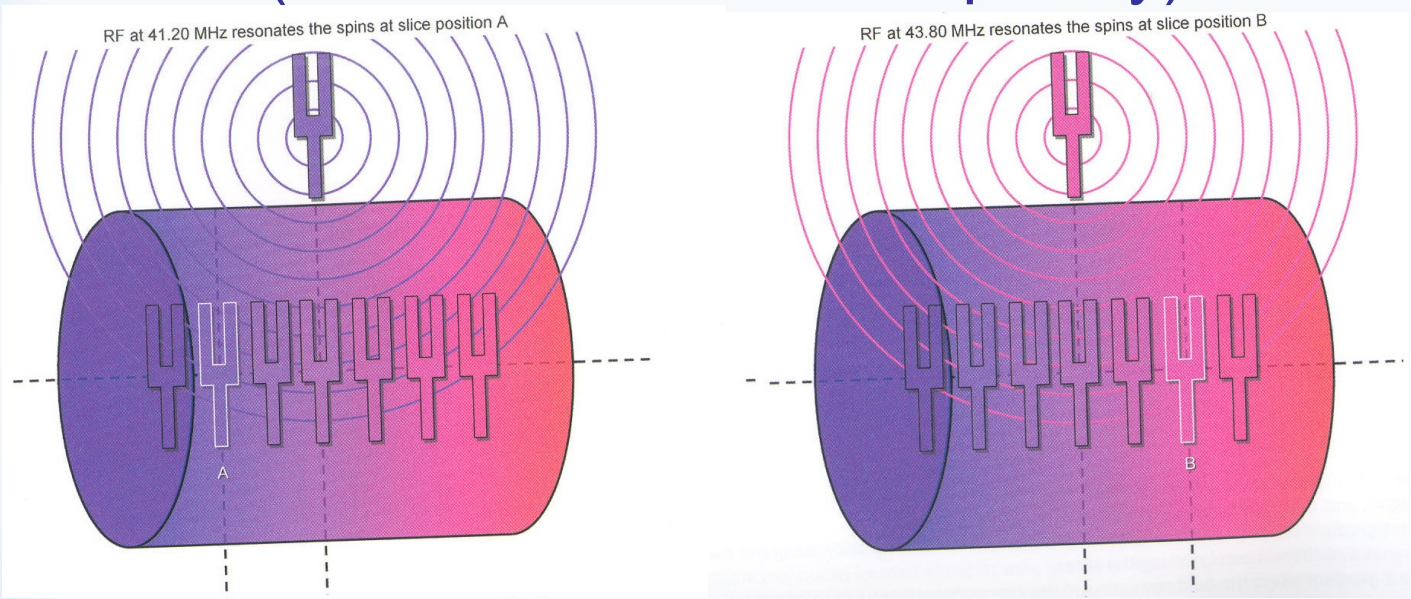


However, gradient echo sequences can be really fast, when short TR and TE times are used.

Actual imaging sequence requires gradients anyway. We will take an spin-echo sequence as an example.



Effect of gradient
(remember Larmor frequency)



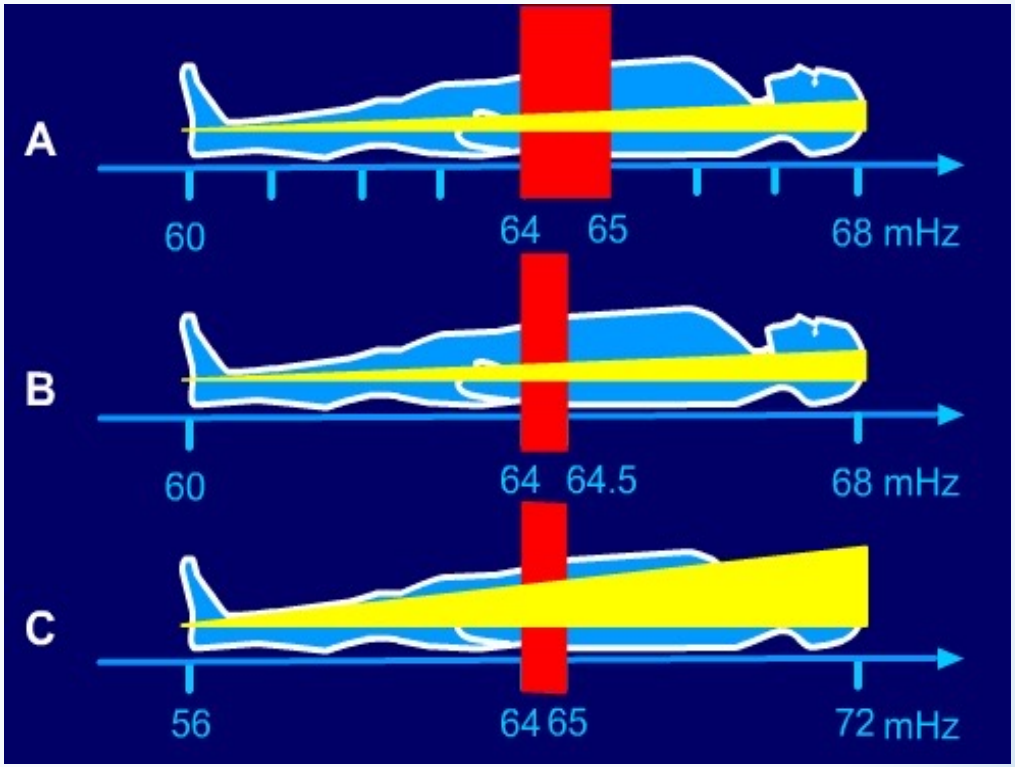
Localizing spins

Slice thickness defined by strength of slice selecting gradient and *bandwidth* of excitation pulse

Bandwidth = 1 MHz

Bandwidth = 0,5 MHz

Bandwidth = 1 MHz
+
Stronger SSG



RF pulse shape

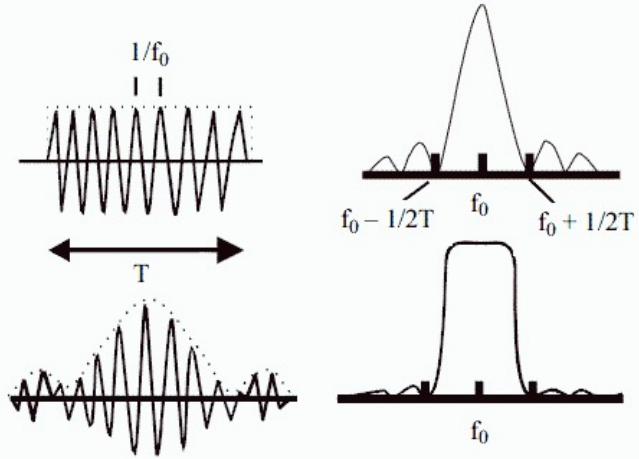
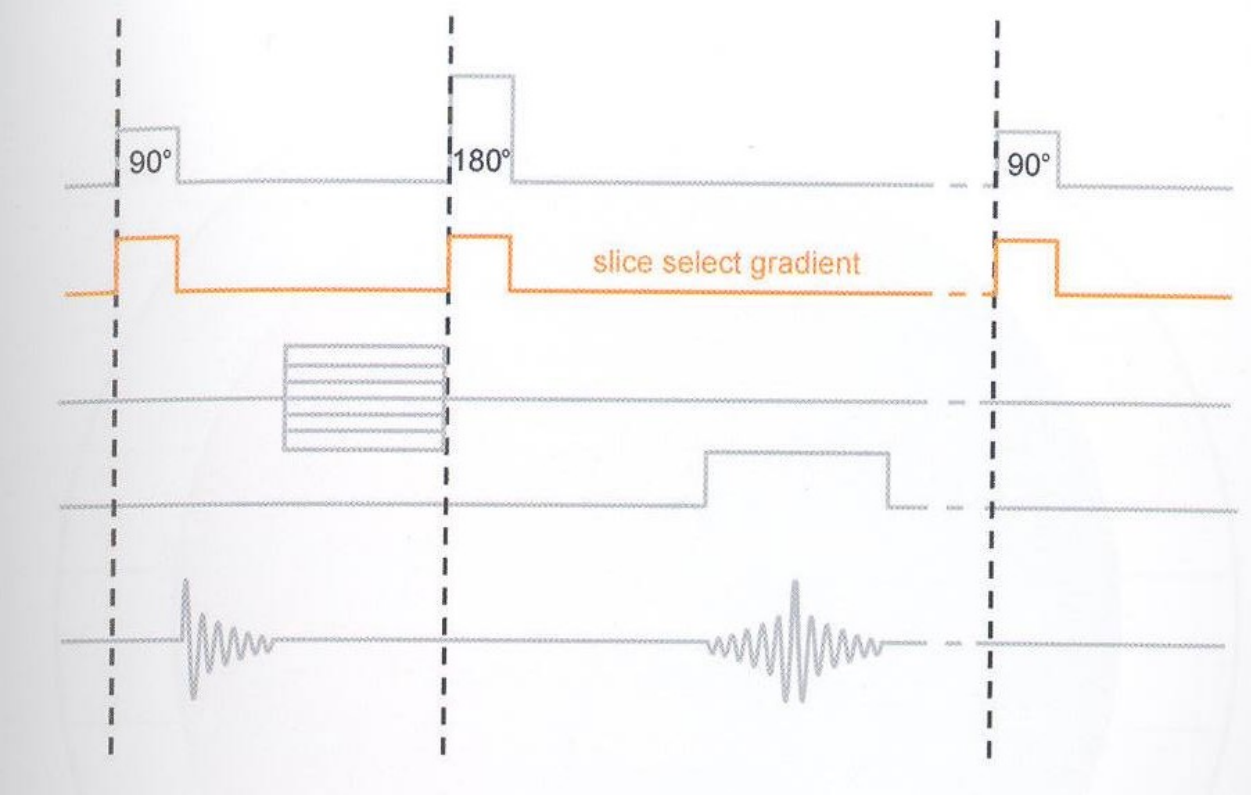
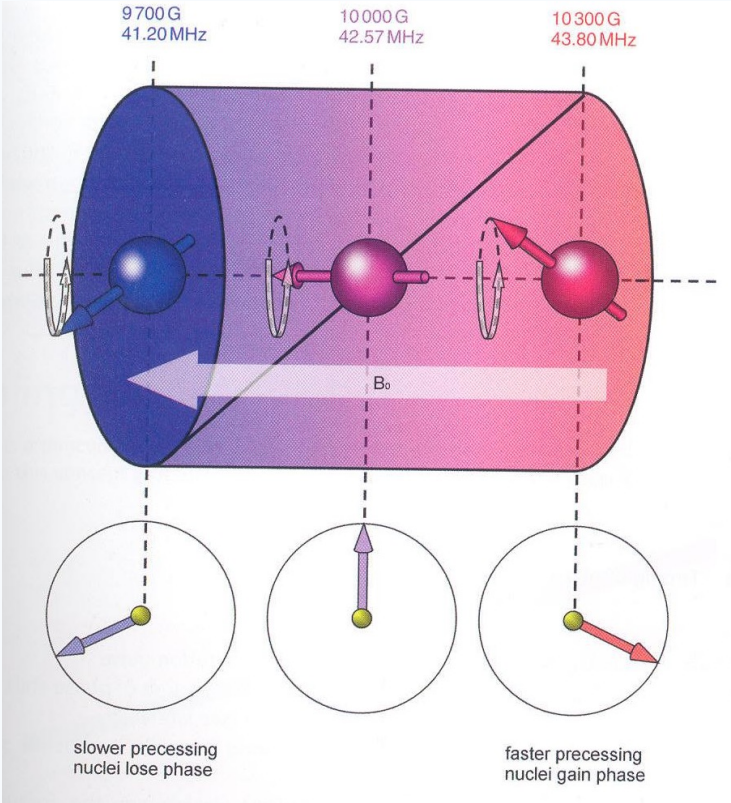


Figure 6.9 A pulse of pure sine waves lasting T seconds gives rise to a sinc function $(\sin(f)/f)$ frequency spectrum. The main central lobe of this spectrum occupies a range of frequencies, $\Delta f \sim 1/T$. This provides an approximate definition of the bandwidth and, in MRI, the slice width. There are however lower amplitude ripples extending to high frequencies. A pulse whose shape or envelope is 'tailored' in time to approximate the central section of a sinc function, yields a frequency spectrum with a steeper-sided central section and lower amplitude ripples.

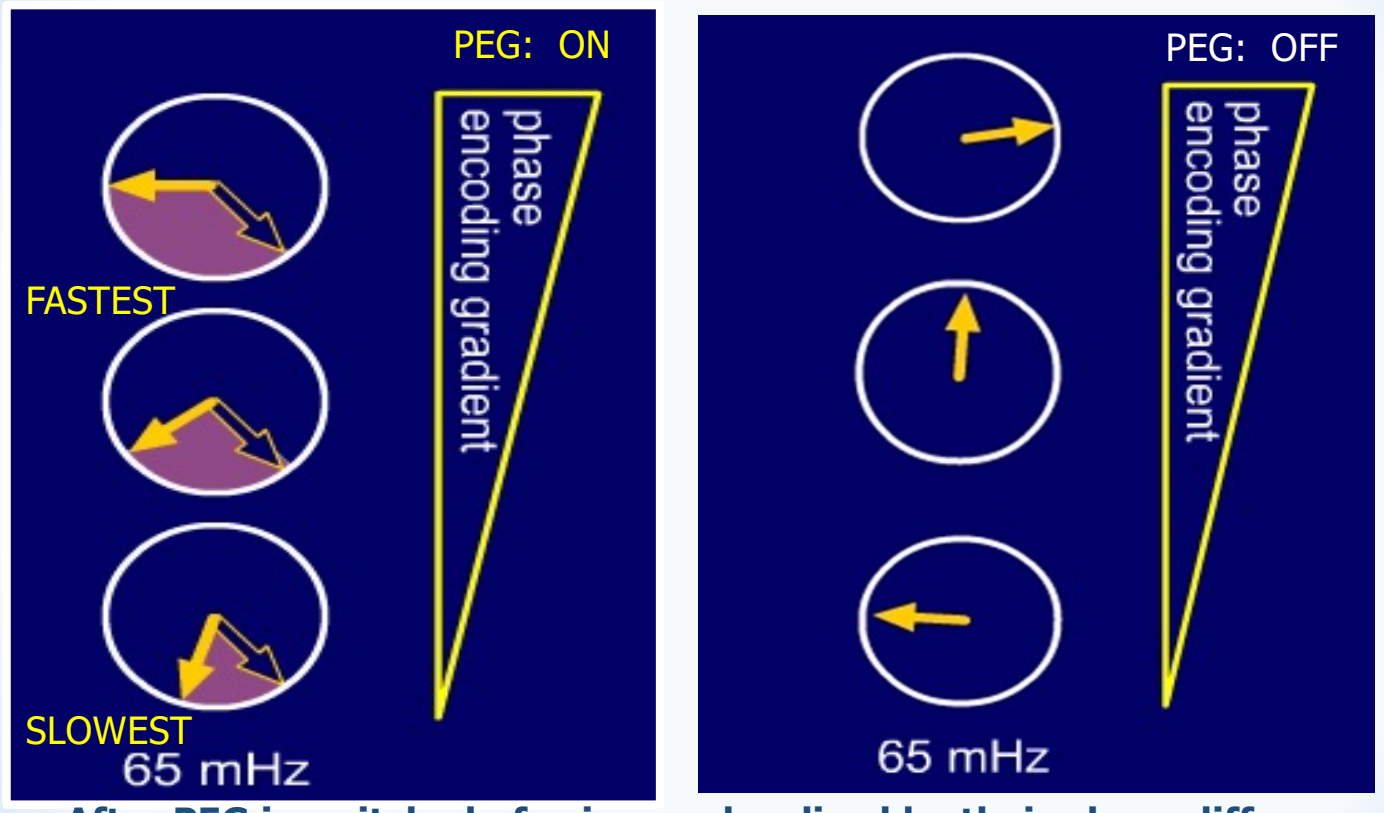
Slice select gradient



Phase encoding gradient

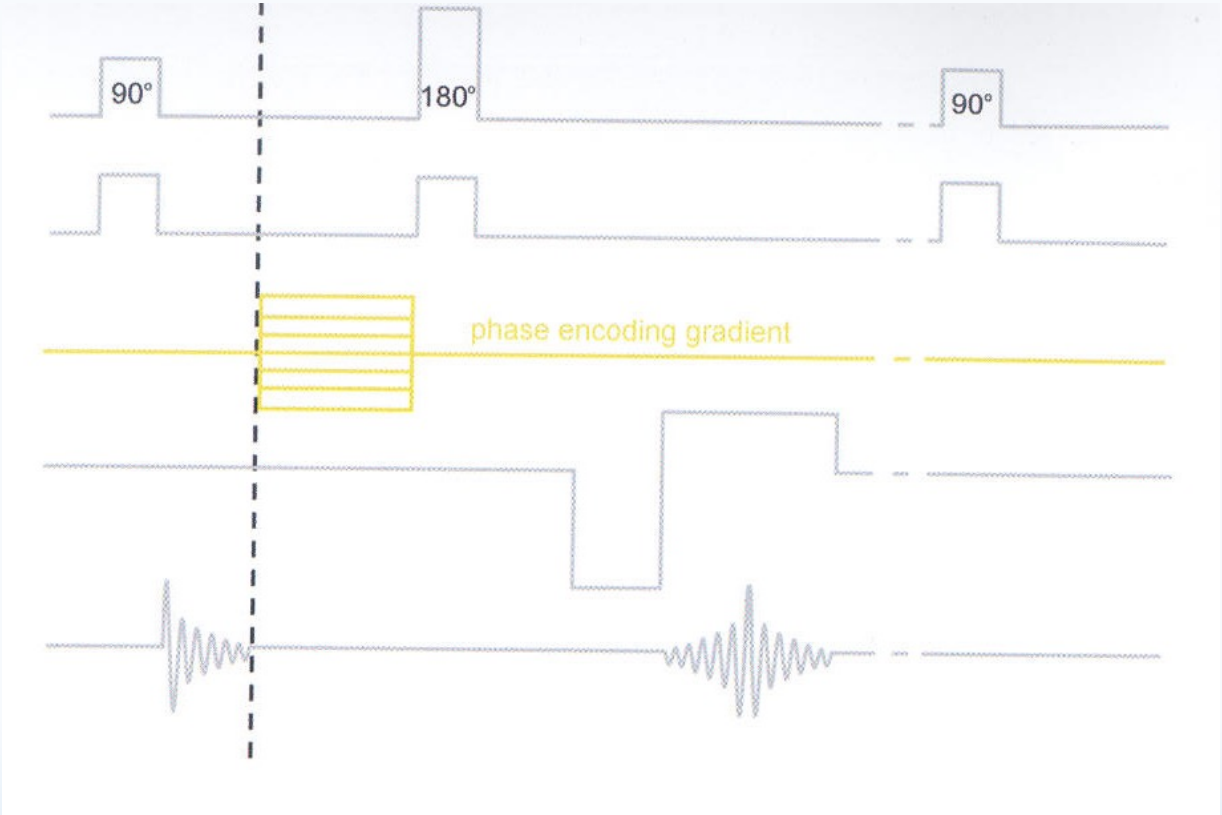


Phase encoding gradient

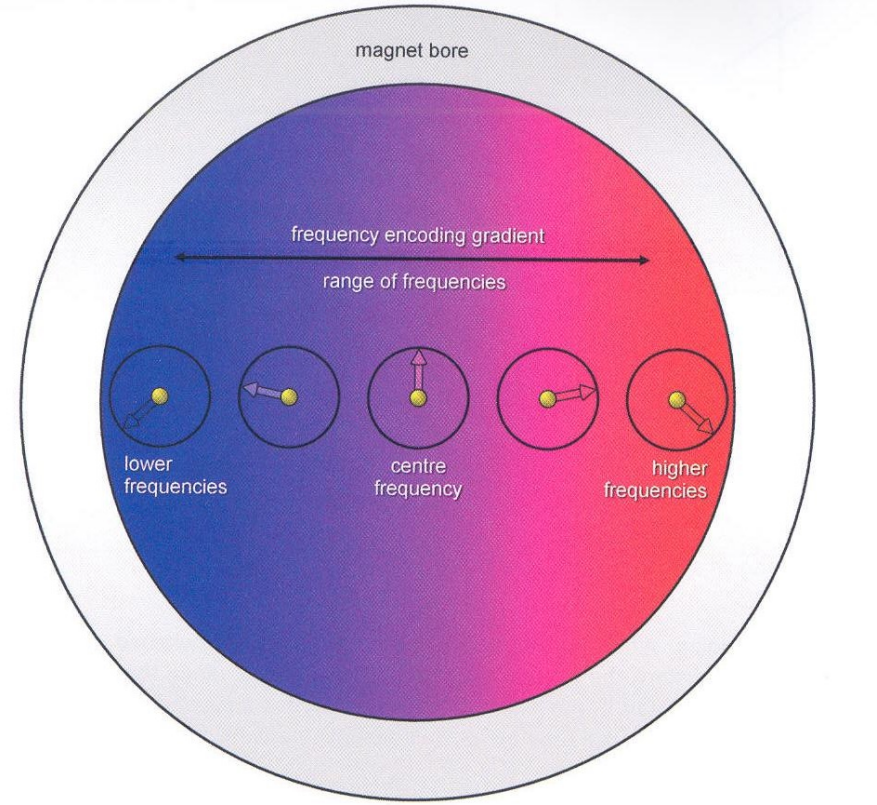


After PEG is switched of spins are localized by their phase difference!

Phase encoding gradient



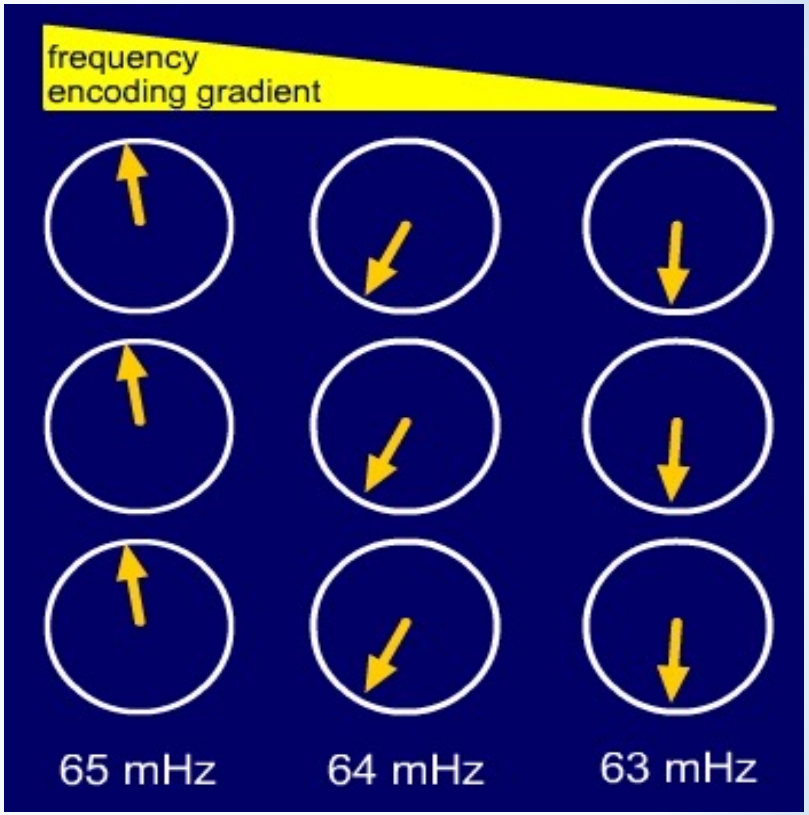
Frequency encoding gradient



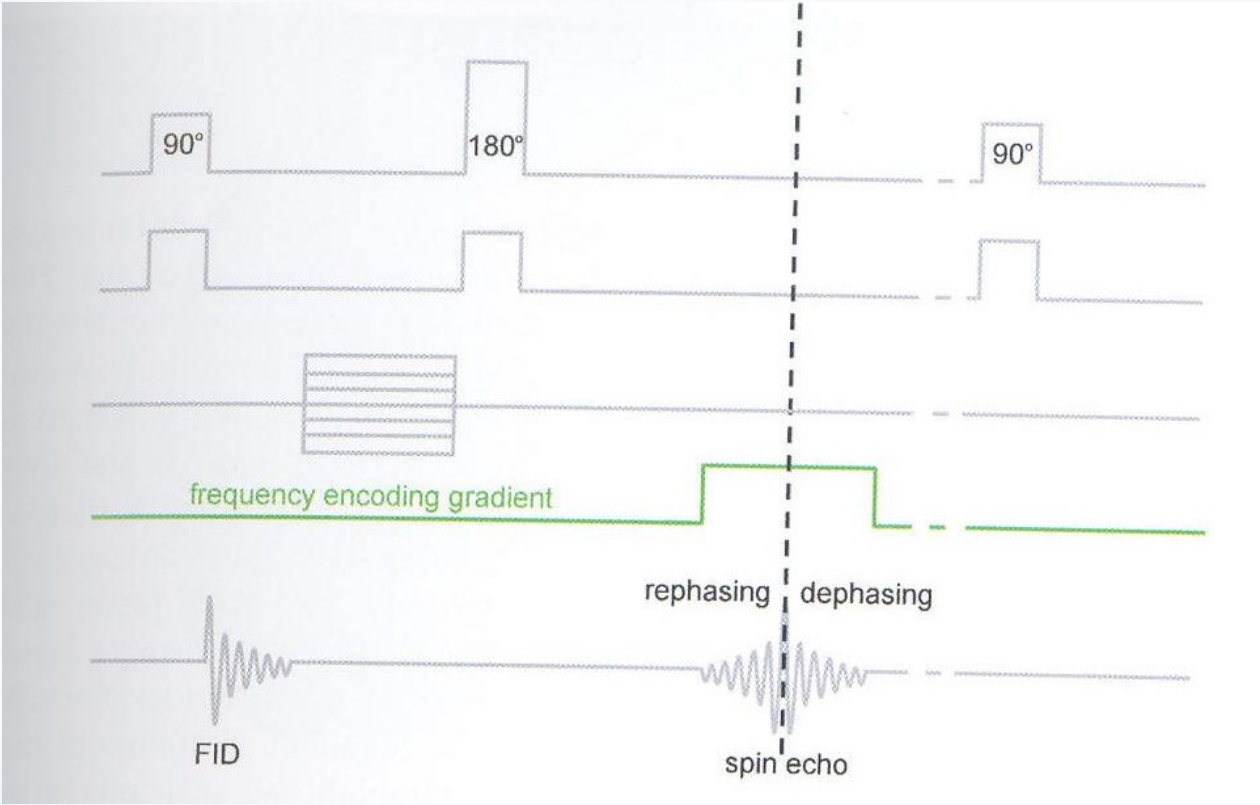
Frequency encoding gradient

After slice is excited and phase encoded, third gradient is applied during the registration of the echo.

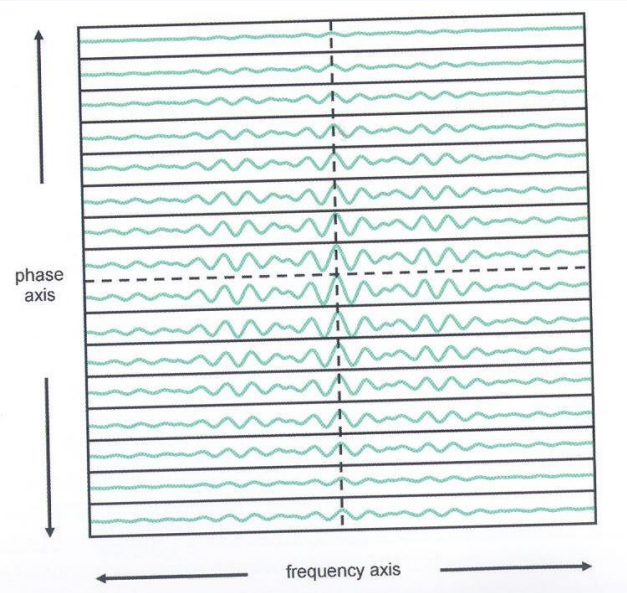
Different frequency during acquisition gives information about the location of the spins.

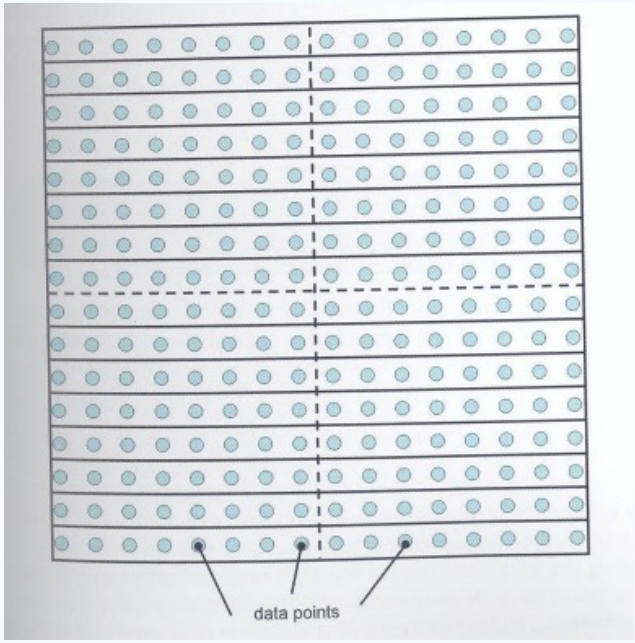


Frequency encoding gradient



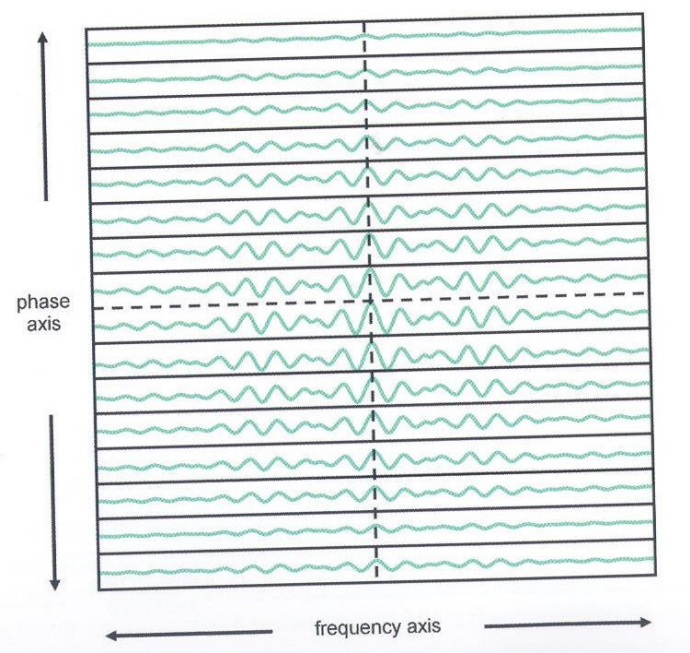
K-space / data points (imaging matrix)

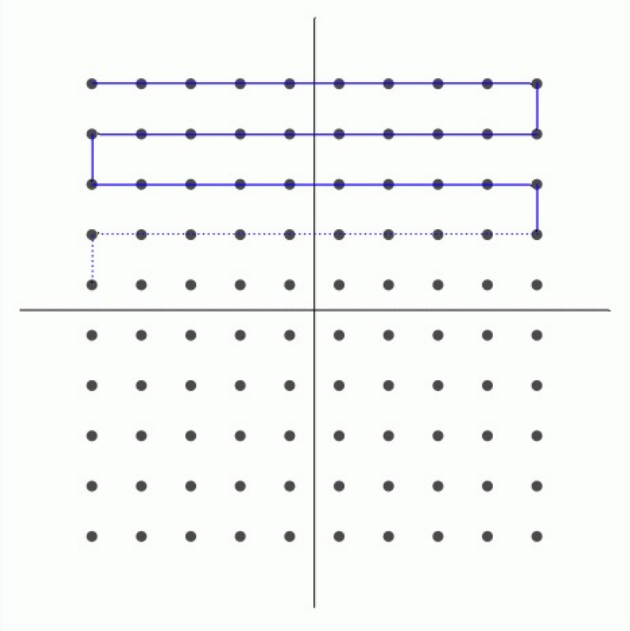




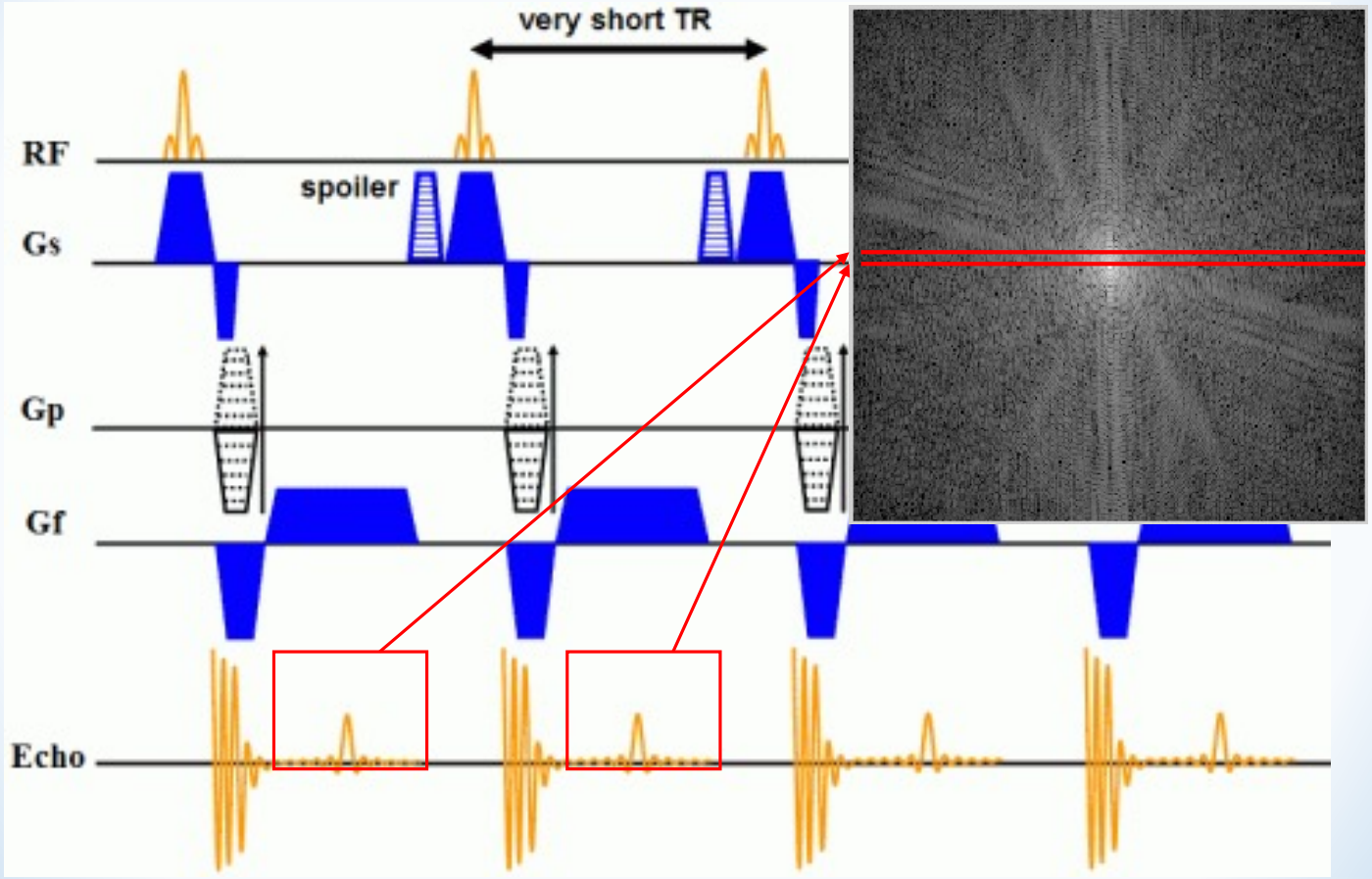
K-space is filled line by line until the whole k-space is covered. The number of datapoint is same as the imaging matrix (for example 256x256).

Acquiring data to k-space





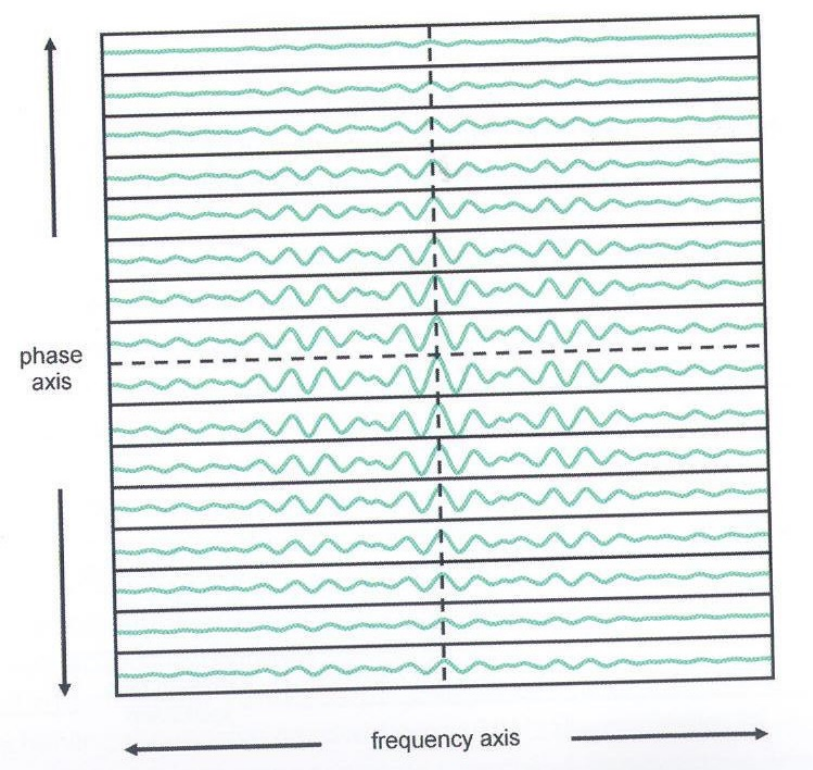
Gradient echo sequence contains exactly same gradients, only 180 degree pulse and it's slice encoding gradient is missing.

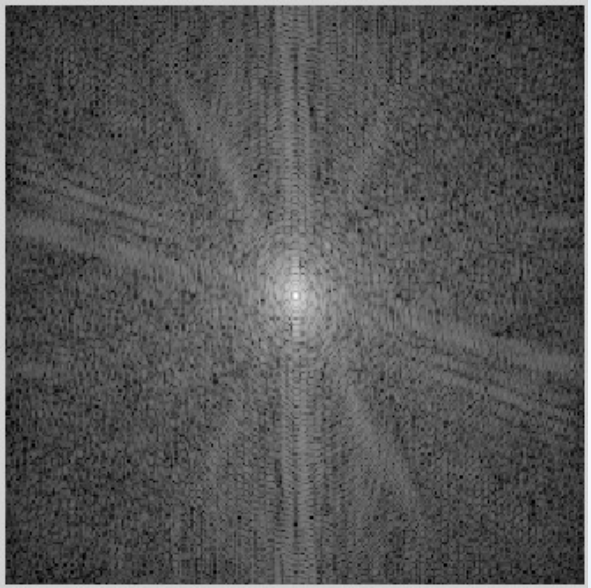


In conclusion, gradients are used for localization of the acquired signal.

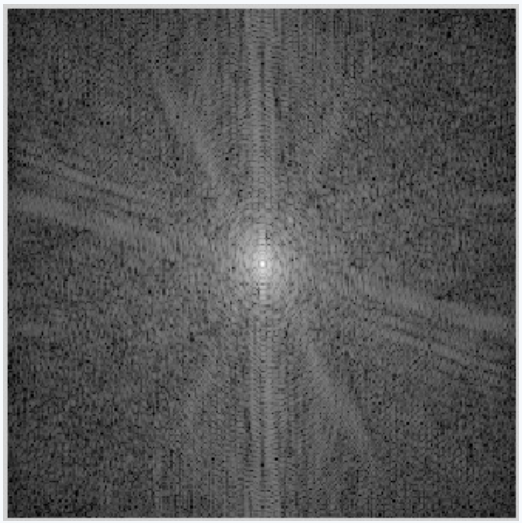
After it is known where the signal is coming from, you can use it to create a map (image) of how strong the signal is in each location.

K-space

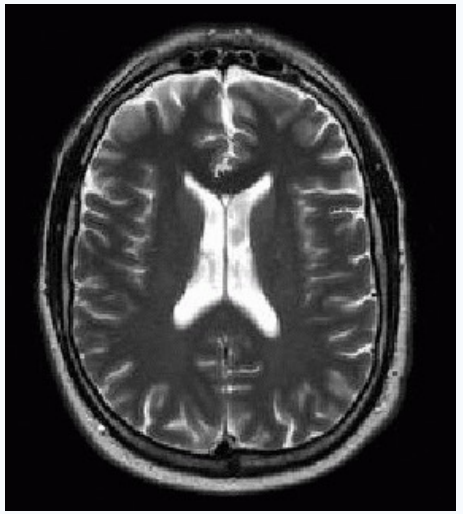




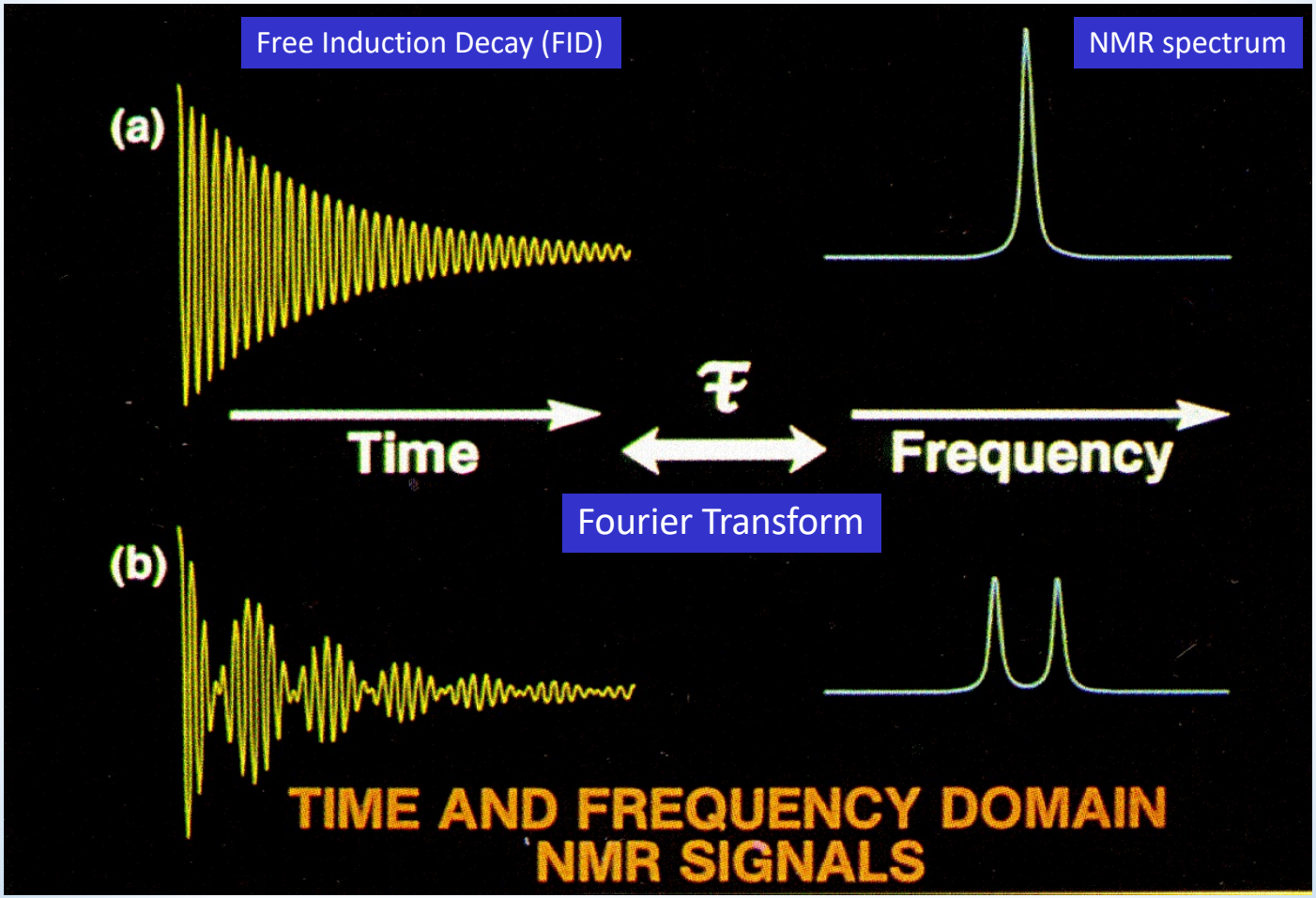
By applying 2D Fast Fourier Transformation, we can get an image from the acquired data of k-space.



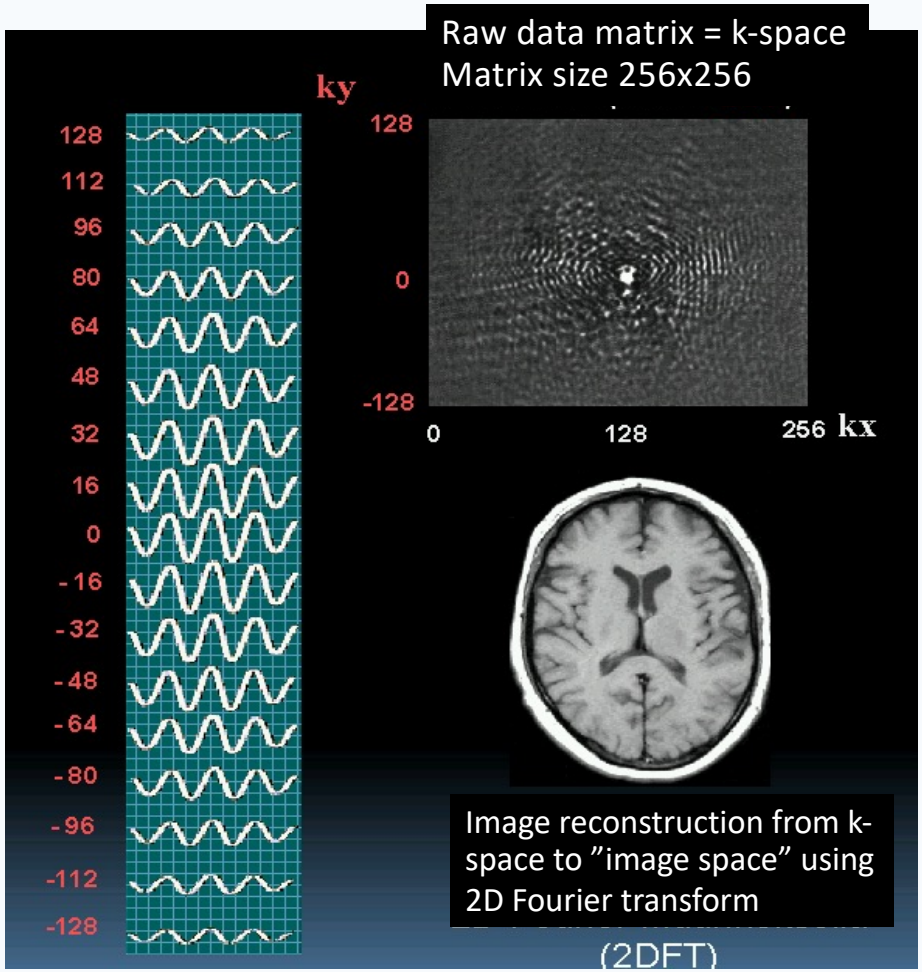
2D FFT

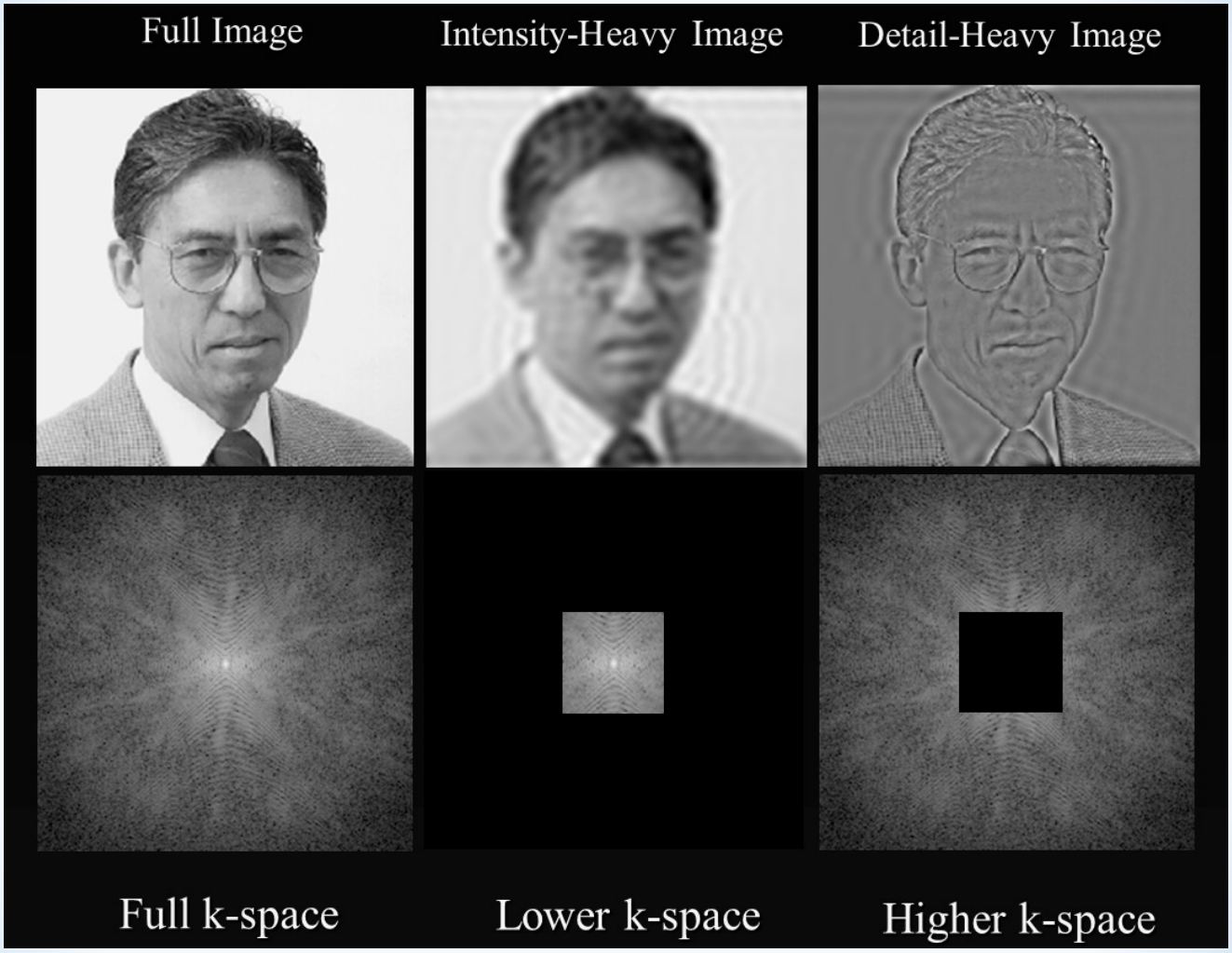


Fourier Transform of FID = NMR spectrum

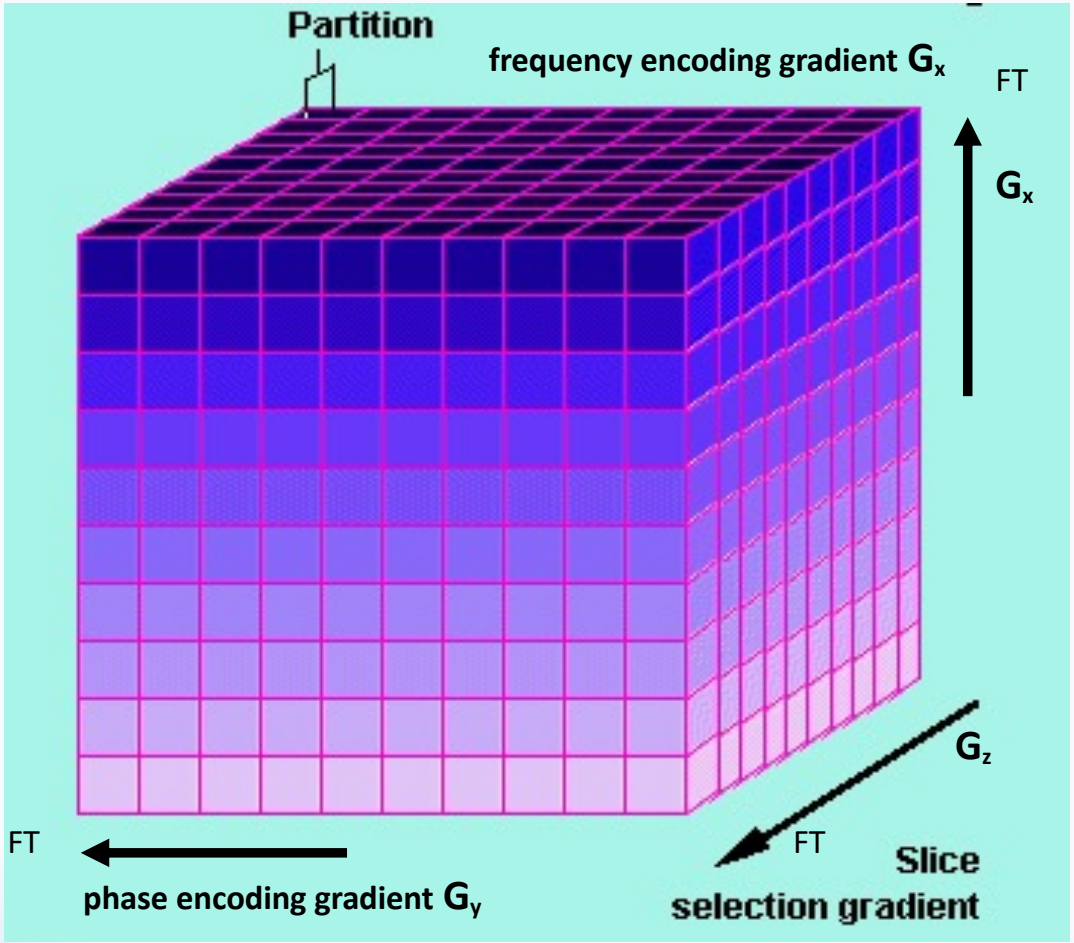


2D- MR image reconstruction




Full Image
Intensity-Heavy Image
Detail-Heavy Image
Full k-space
Lower k-space
Higher k-space

3D- FT MR image reconstruction



MRI From Picture to Proton, 2nd Ed
McRobbie, Moore, Graves, Prince
(2006) Cambridge

3D MP-RAGE sequence

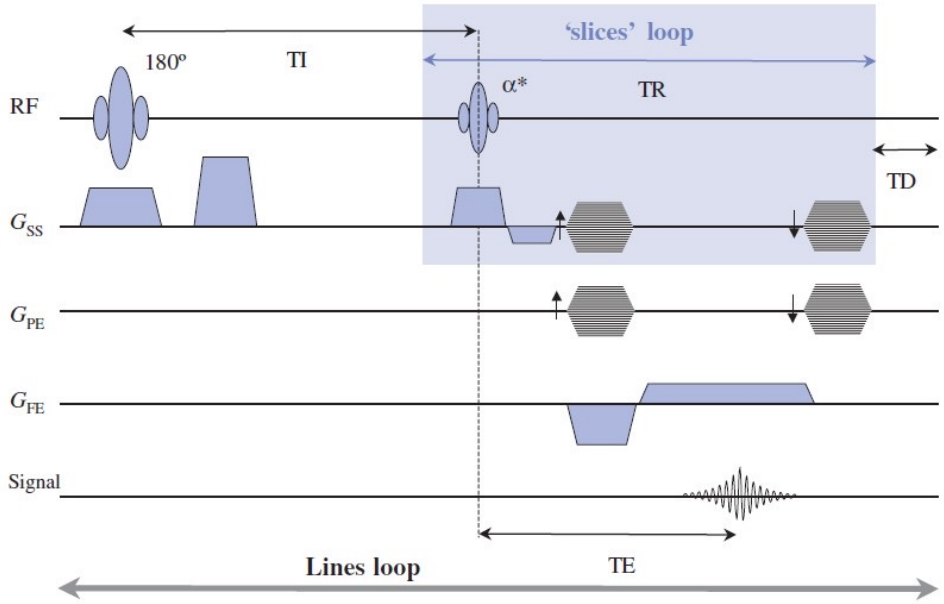


Figure 12.38 3D MP-RAGE sequence. Note that when RF spoiling (denoted by asterisk) is deployed the phase-encode gradients are rewound.

3D Magnetization Prepared Rapid Acquisition by Gradient Echo

MRI From Picture to Proton, 2nd Ed
McRobbie, Moore, Graves, Prince
(2006) Cambridge

3D MP-RAGE sequence

MP-RAGE or Magnetization Prepared Rapid Acquisition by Gradient Echo is the same in principle as turbo-FLASH. However, the name has tended to apply to a particular 3D implementation of turbo-FLASH. As a 3D technique there are too many combinations of the two phase-encode gradients to acquire the whole of 3D k-space from a single preparation. The solution is to acquire all the 'slice' or in-depth lines of data from each prep, then introduce a recovery delay (as in segmented turbo-FLASH) before moving on to the next in-plane line of k-space. As a result the in-plane resolution is not compromised, the data being all acquired with the same degree of relaxation.

MP-RAGE can produce very high resolution, T1-weighted images showing very good anatomical detail, particularly of the brain . The introduction of this delay means that MP-RAGE is not ultra-fast in its scan time (although it uses ultra-fast methods).

3D MRI

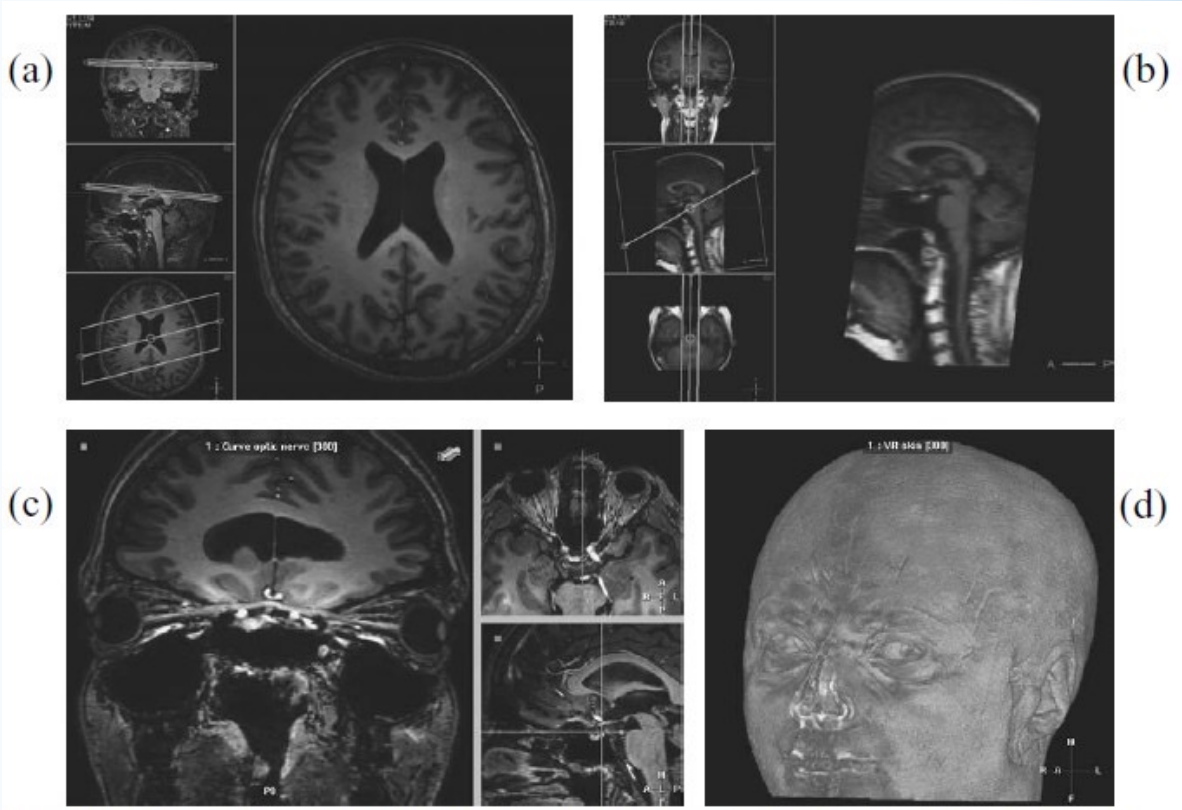


Figure 4.16. (a) Oblique slice reformatted from a 3D volume. (b) The "staircase" effect of having non-isotropic voxels. (c) Curved reformat of the optic nerve. (d) Shaded surface display. Note the crows-feet around the eyes!

MRI From Picture to Proton, 2nd Ed
McRobbie, Moore, Graves, Prince (2006) Cambridge

Reconstruction of MR image

“Probably every digital medical image produced, now involves some use of the FFT; if not in tomographic reconstruction, then in filtering or correlation analysis.”

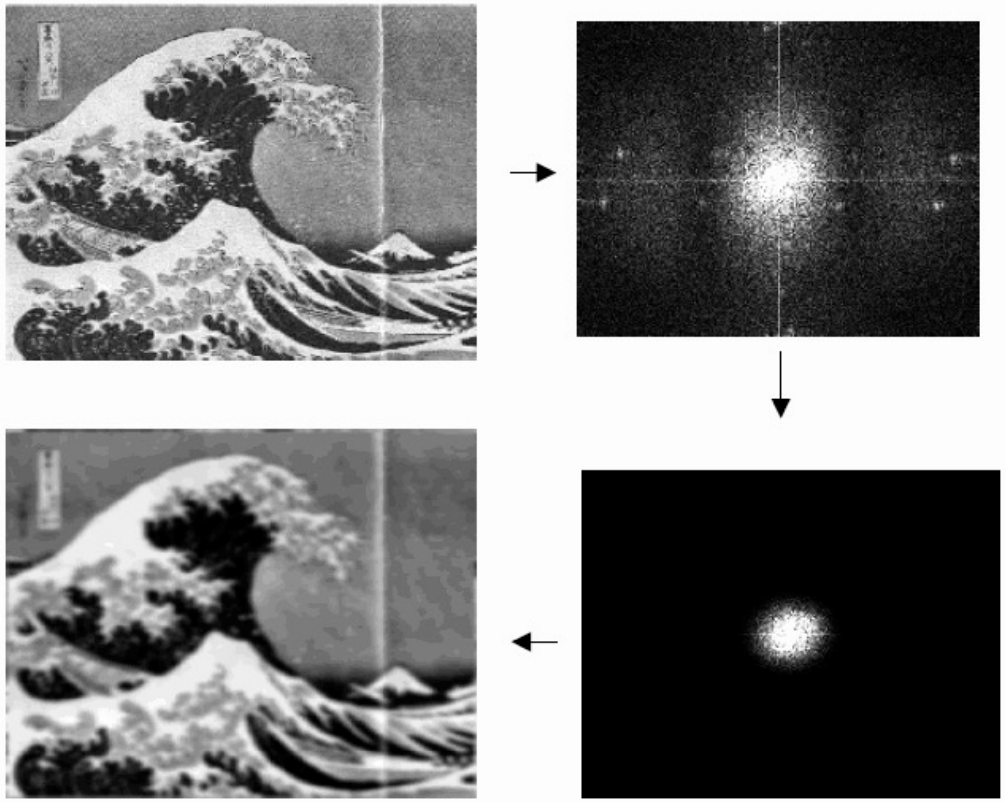


Figure 1.4 The monochrome Hokusai, in the top left panel, is Fourier transformed to produce its *K*-space map in the top right-hand panel. The *K*-space map is filtered, to reduce the amplitudes of the larger *K* values (shorter waves) in the bottom right-hand panel. The filtered *K*-space waves are then recombined to produce the filtered (here blurred) image in the bottom left-hand panel. Here we have obliterated all but a tiny central portion of the *K*-space map (all short waves) but still the reconstructed image is still easily recognisable as the Hokusai.

An Introduction to the Fourier Transform: Relationship to MRI

Thomas A. Gallagher[1]
Alexander J. Nemeth[1,2]
Lotfi Hacein-Bey[1]

OBJECTIVE. The Fourier transform, a fundamental mathematic tool widely used in signal analysis, is ubiquitous in radiology and integral to modern MR image formation. Understanding MRI techniques requires a basic understanding of what the Fourier transform accomplishes. MR image encoding, filling of k-space, and a wide spectrum of artifacts are all rooted in the Fourier transform.

CONCLUSION. This article illustrates these basic Fourier principles and their relationship to MRI.

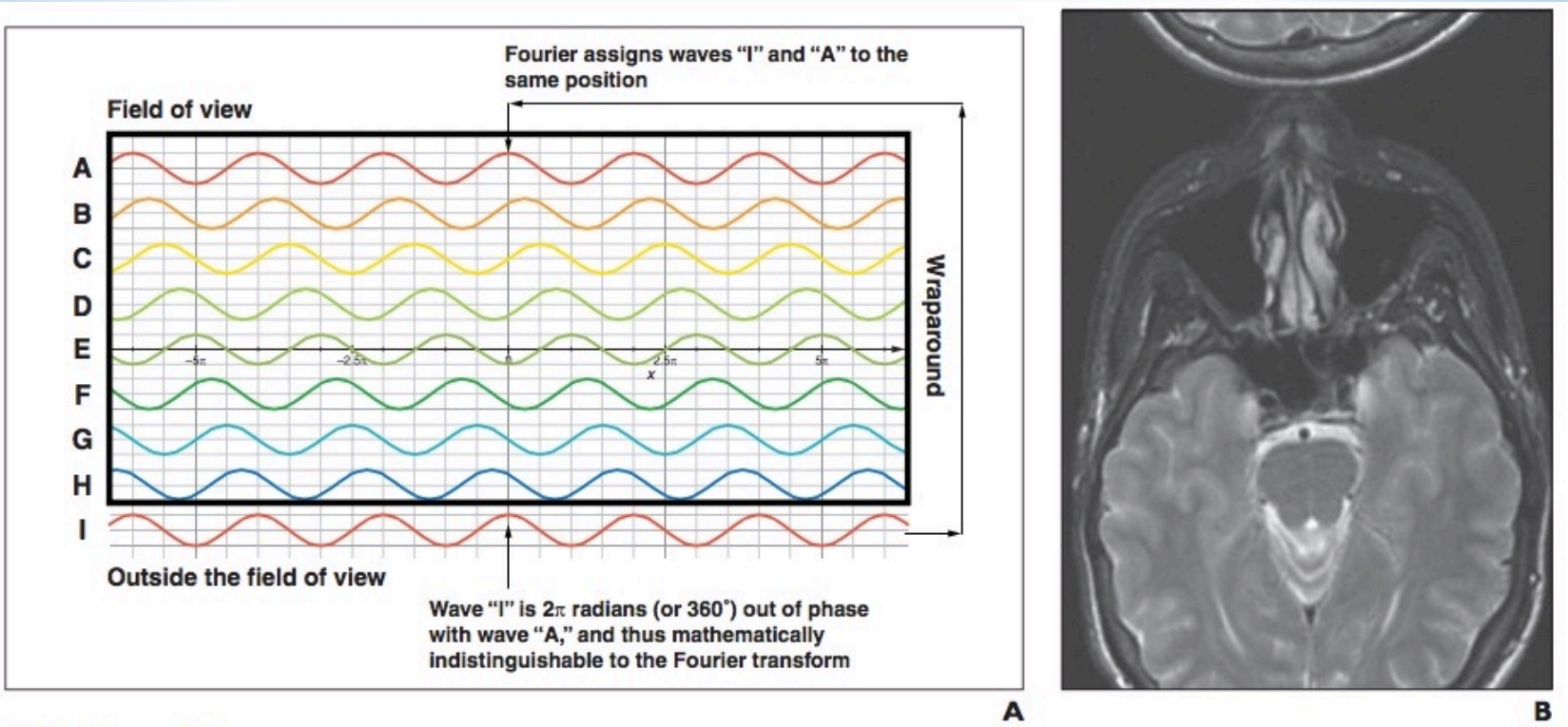
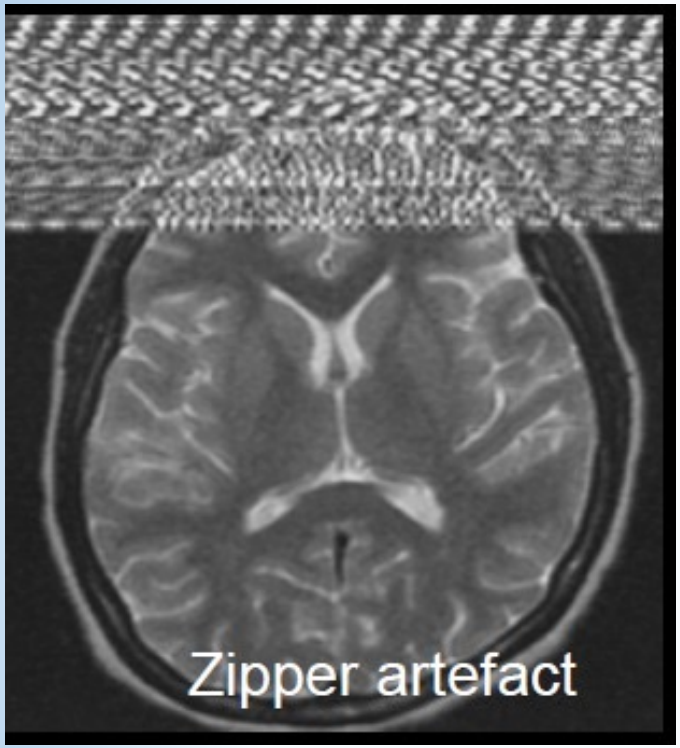
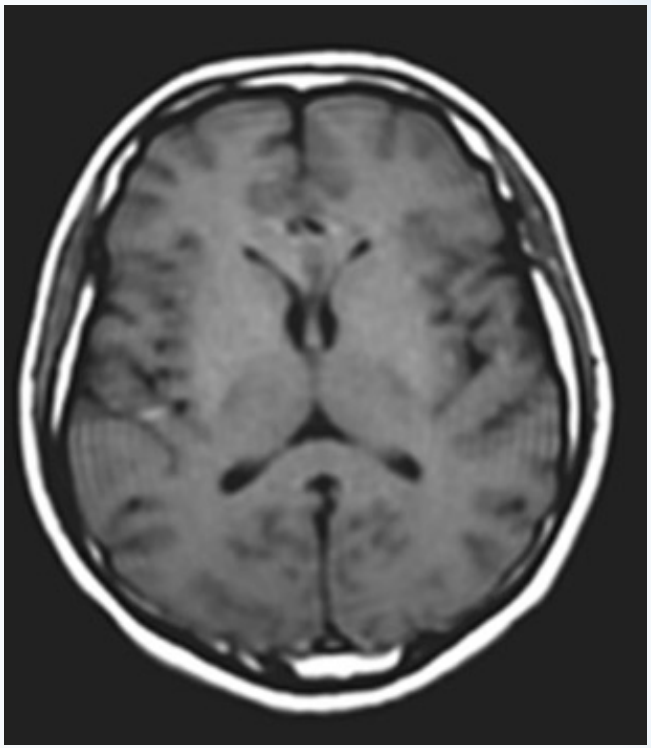


Fig. 12—Wraparound.
A, Wraparound (aliasing). Only phase shifts between 2π radians or 360° are available to encode an image. The phase shift in this image (depicted by waves A–H) covers the field of view (*black rectangular outline*). Just outside the field of view, wave "I" has assumed a phase shift of 2π radians (360°) and is mathematically identical to wave "A" on the opposite side of the field of view. Fourier transform assigns structures encoded by "I" to positions encoded by "A," giving the wraparound phenomenon.
B, Axial T2-weighted image shows back of head (excluded from field of view) wrapping around to the front.

Gallagher et al AJR (2018) Intro to Fourier Transform

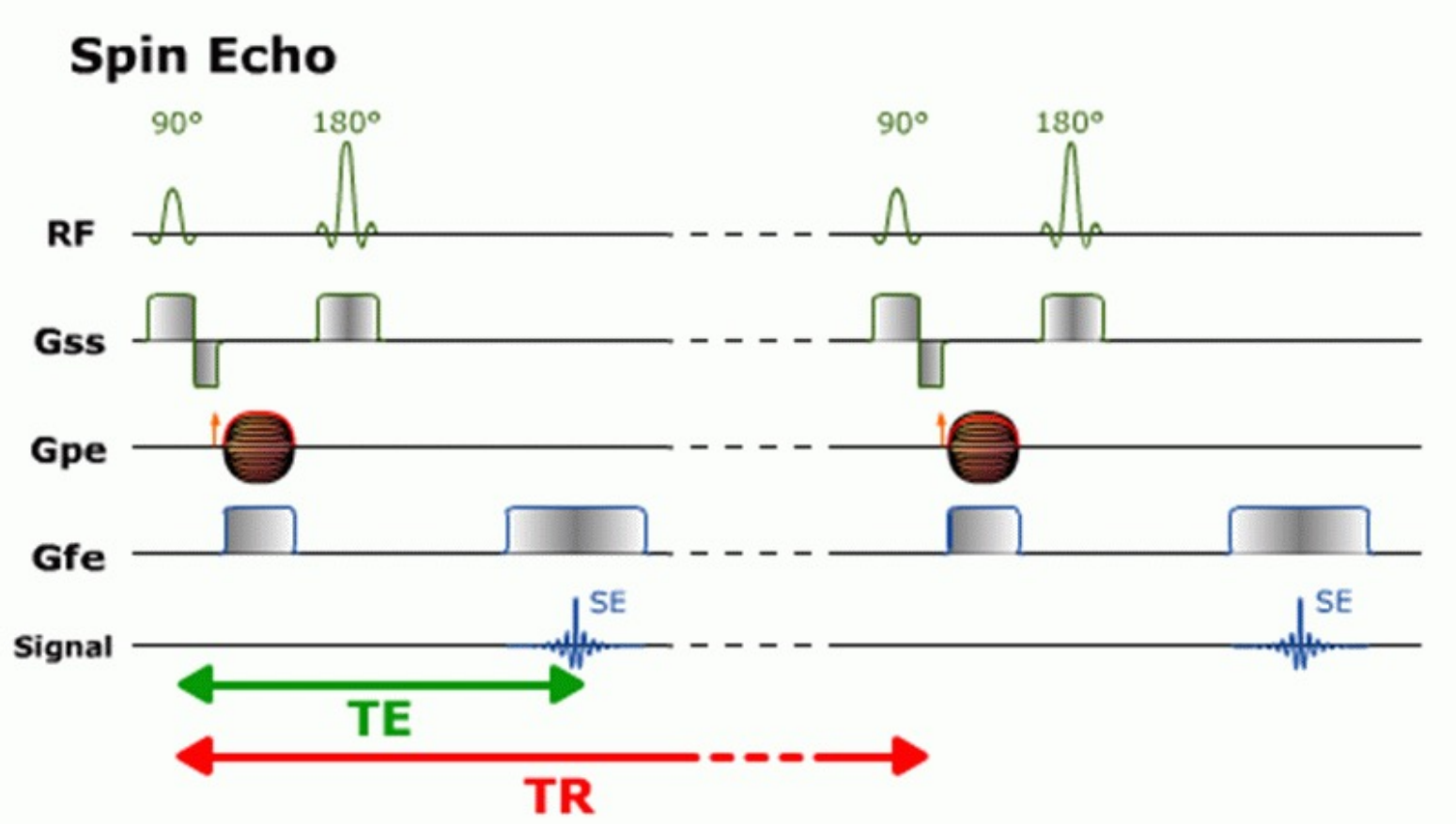




Gibbs ringing artefact

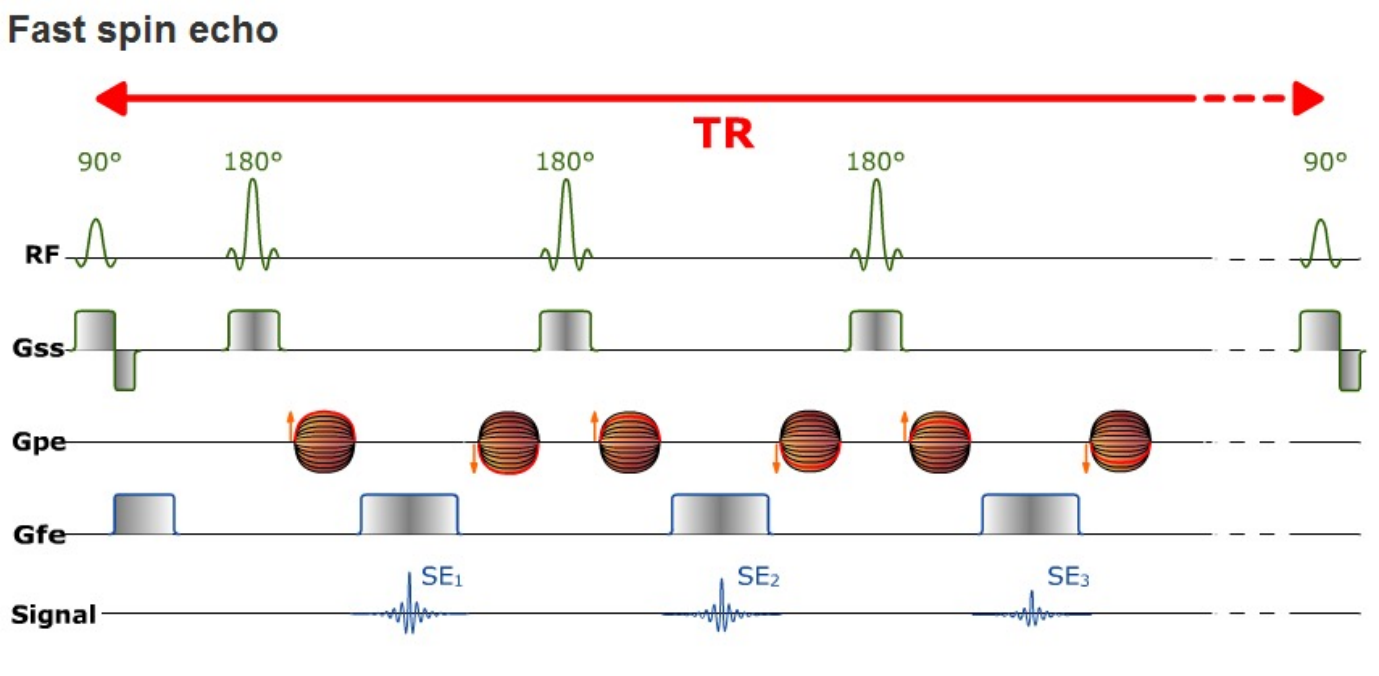
Sequences

Spin Echo sequence

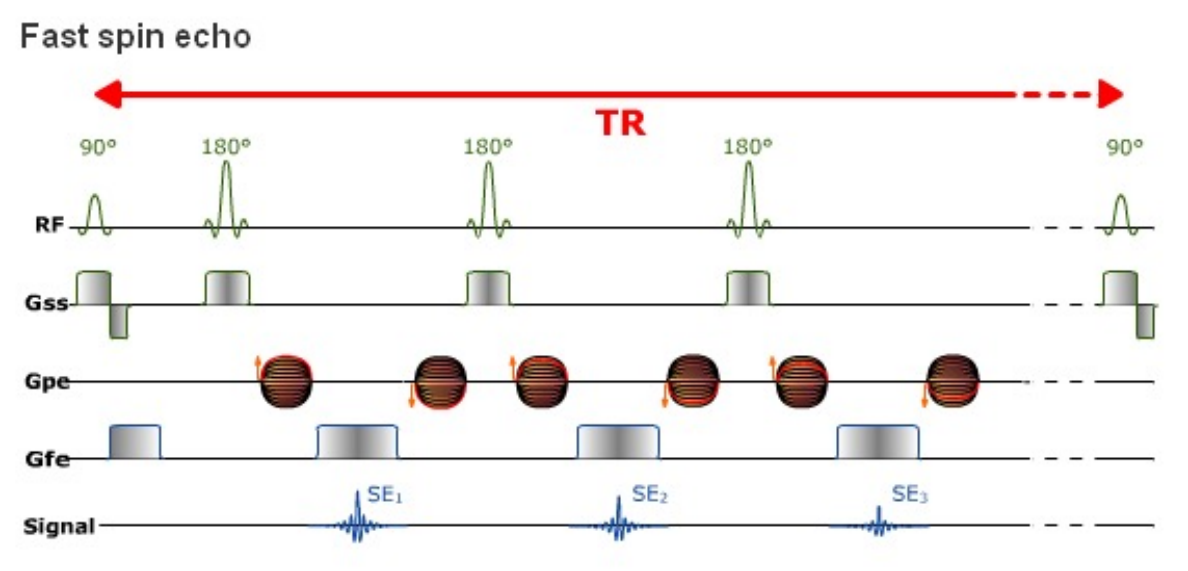


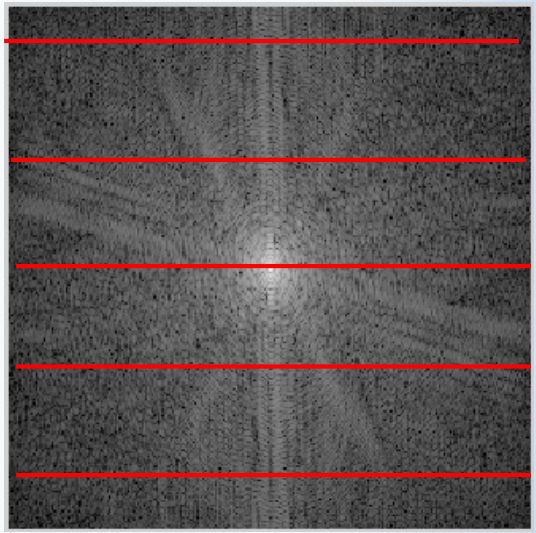
Fast Spin Echo “FSE” sequence (Turbo Spin Echo “TSE”)

FSE technique is based on several phase encoding steps after each excitation. This is possible by applying opposite phase encoding gradient after the echo and then adding 180 degree pulse to create a new echo. Using different phase encoding gradient than during the first echo, new line can be collected to k-space.
This can be repeated several times (more than 20 times in T2-weighted sequences).



Within one TR several echoes are collected to k-space





Contrast in FSE is based on “effective TE”

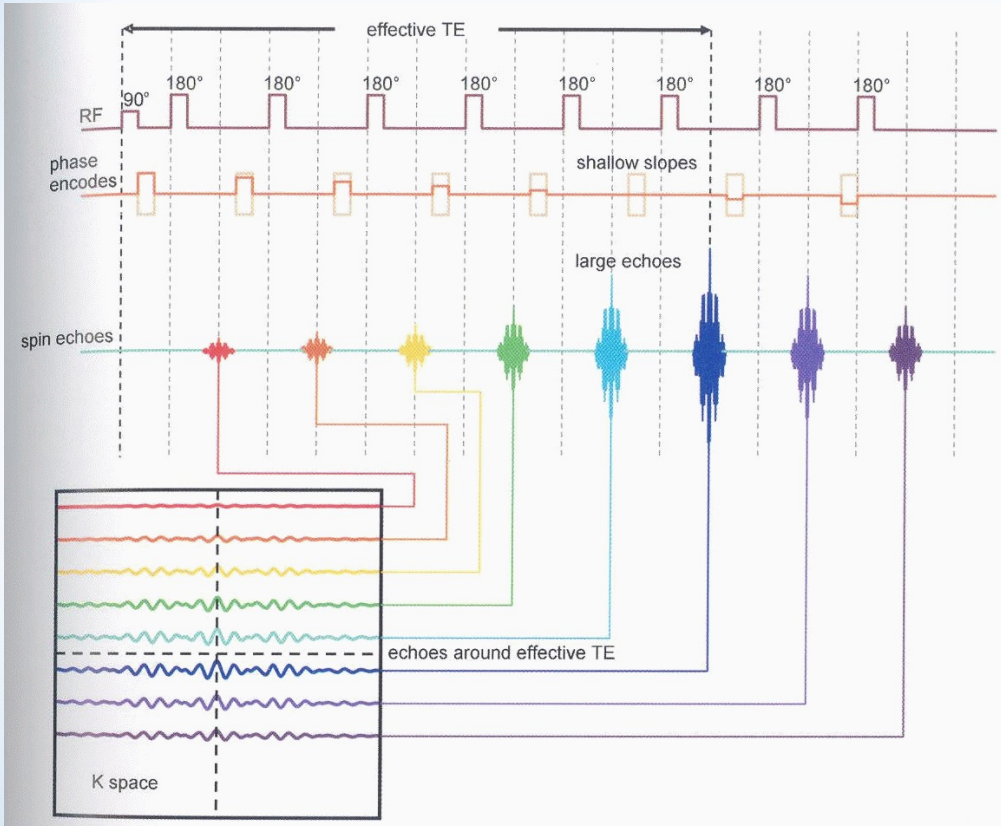


Figure 5.5 K space filling and phase re-ordering.

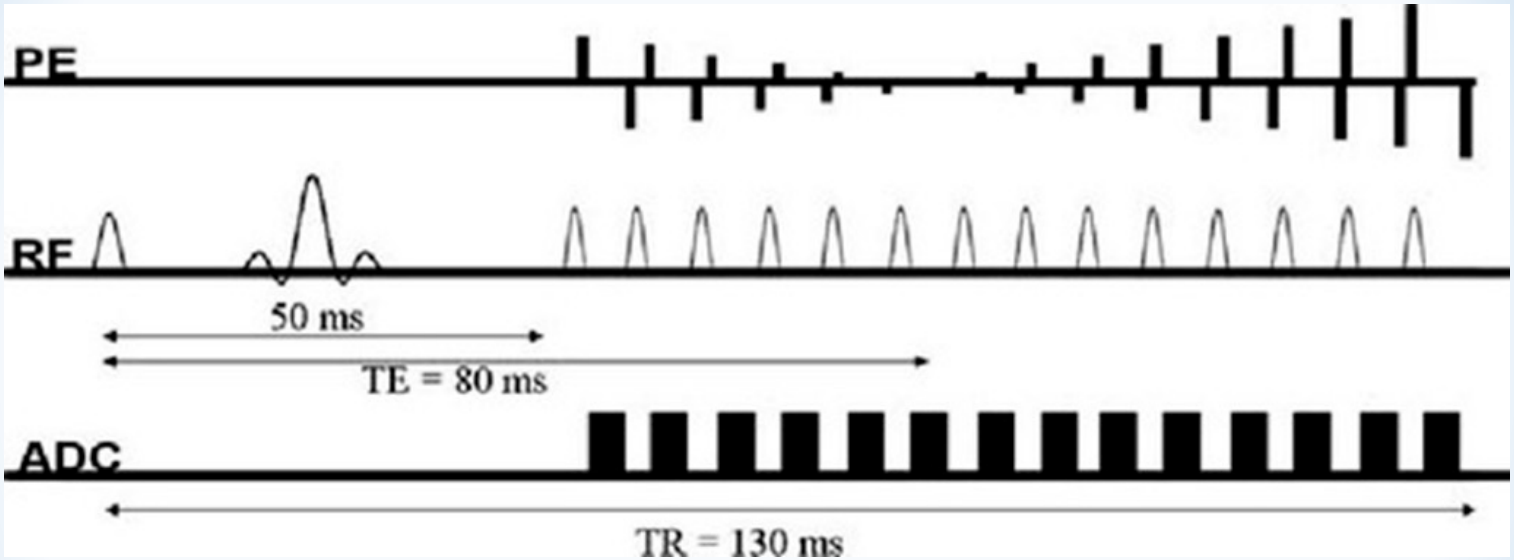
Central part of k-space is defining the contrast of image. Therefore, echoes that have required TE are collected to the middle of k-space.

For example:
In PD sequence echoes with 10-30 ms TE are in the center of k-space

In T2 echoes with 80-120 ms are in the middle

HASTE (Half Fourier Acquisition Single Shot Turbo Spin Echo)

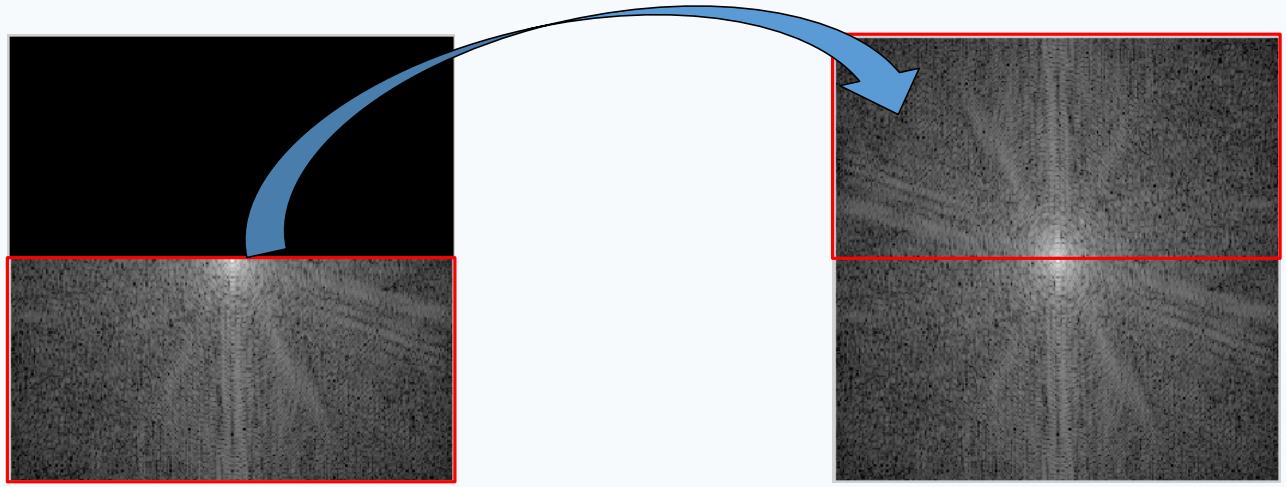
All the k-space lines are acquired after single excitation (sigle-shot imaging)



Resolution is often poor, but it is possible to scan even moving objects.

HASTE (Half Fourier Acquisition Single Shot Turbo Spin Echo)

In addition HASTE-sequence is taking advantage of symmetry of k-space.



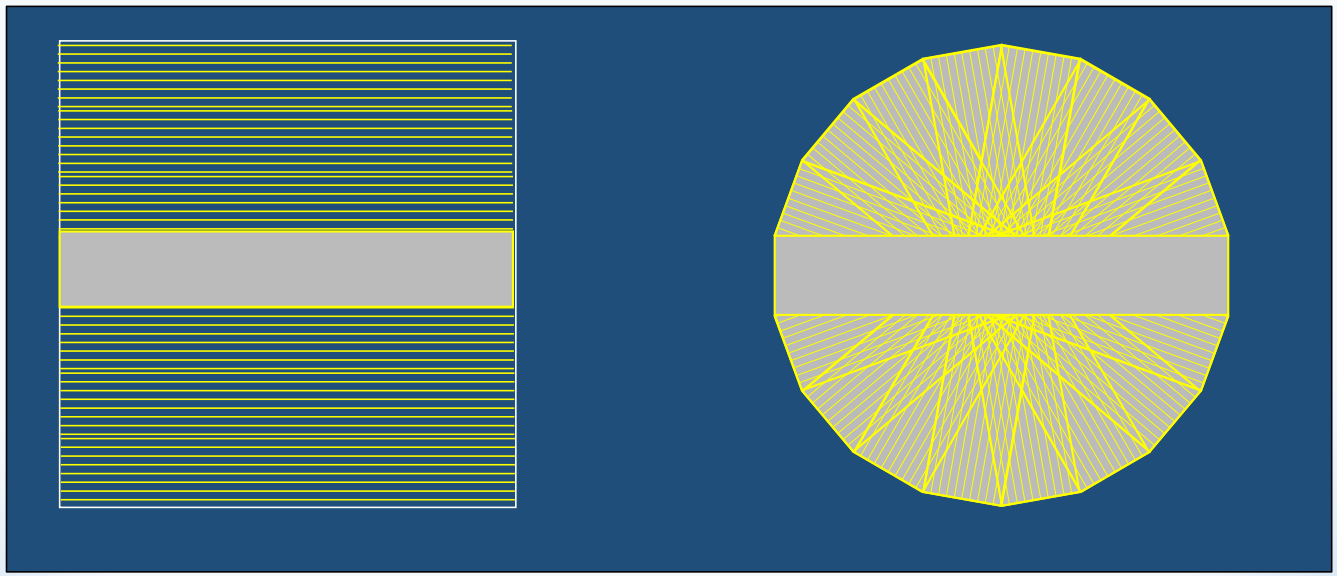
Provided no phase errors occur during data collection, k-space possesses a peculiar mirrored property known as conjugate (or Hermitian) symmetry. So, in theory, it is possible to create image using only half of the k-space data.

Special techniques

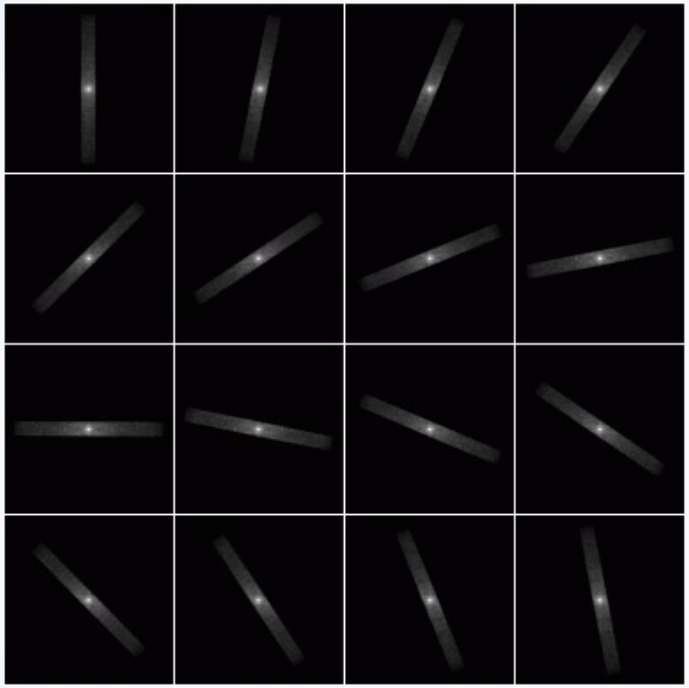
PROPELLER

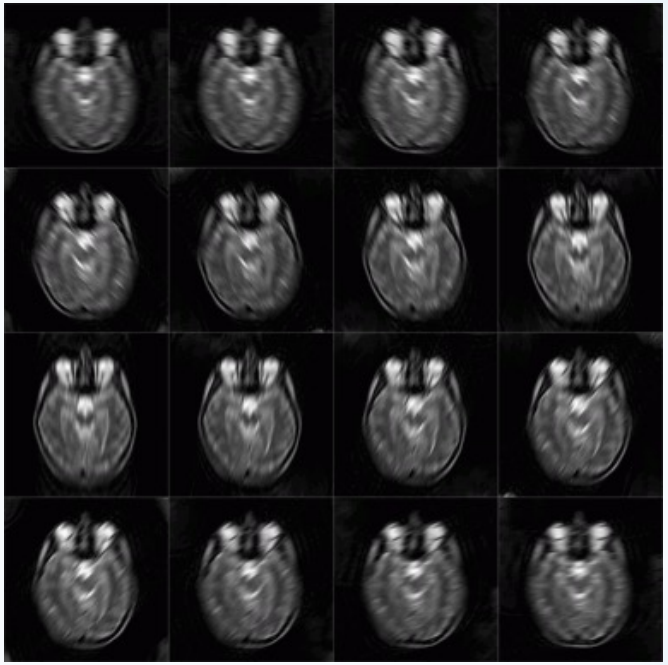
(**P**eriodically **R**otated **O**verlapping Parall**EL** **L**ines with **E**nhanced **R**econstruction technique was developed in the late 1990s as a motion reduction method. The basic idea was to sample *k*-space in a rotating fashion using a set of radially directed strips or "blades".

Each blade is composed of multiple parallel phase-encoded lines that can be collected using fast spin echo or gradient echo methods. In common practice, 8-32 blade lines are acquired in a single shot. The blades are then rotated by a small angle (10°−20°) at which time a second set of data are acquired. The process continues until imaging data from the entire *k*-space circle has been collected.

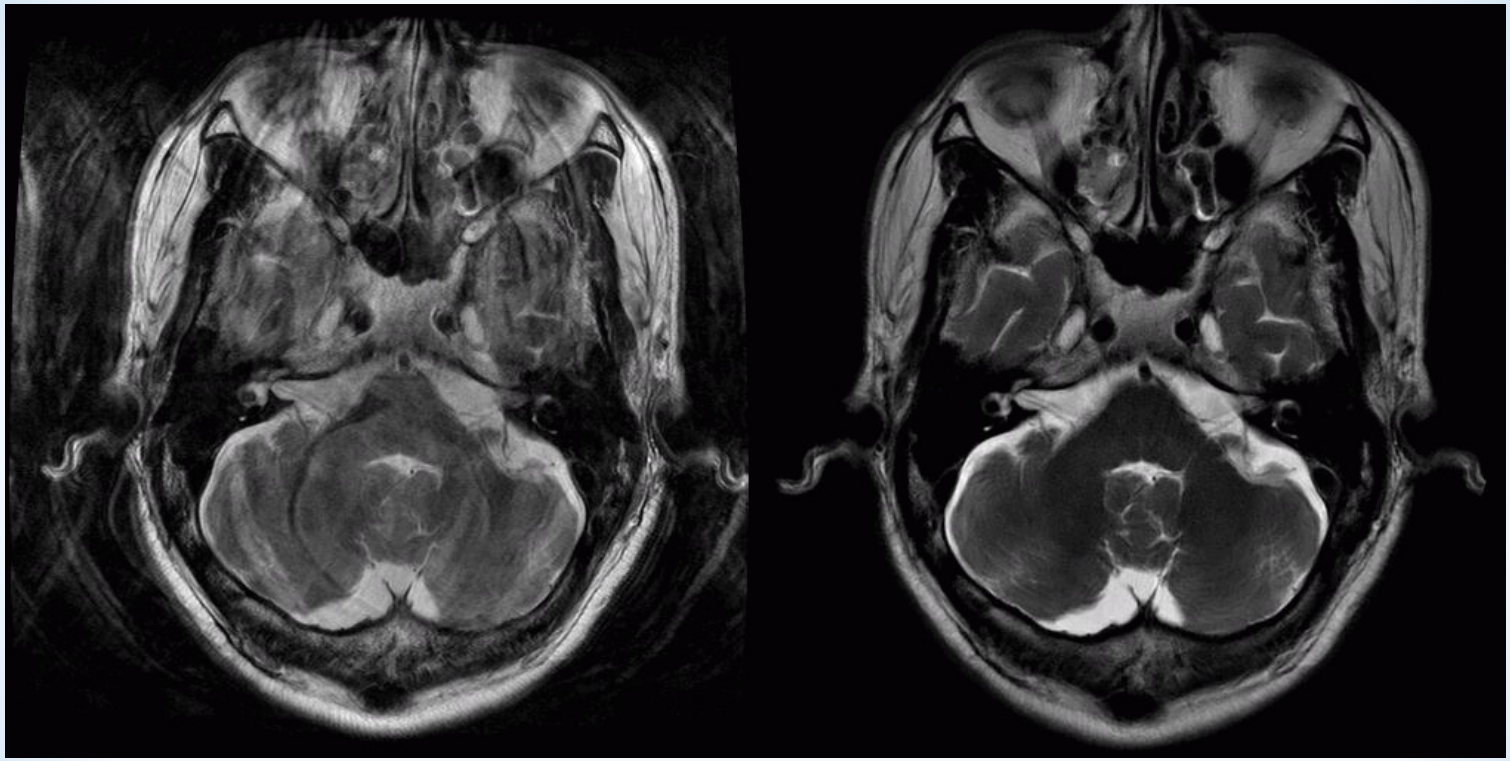


PROPELLER





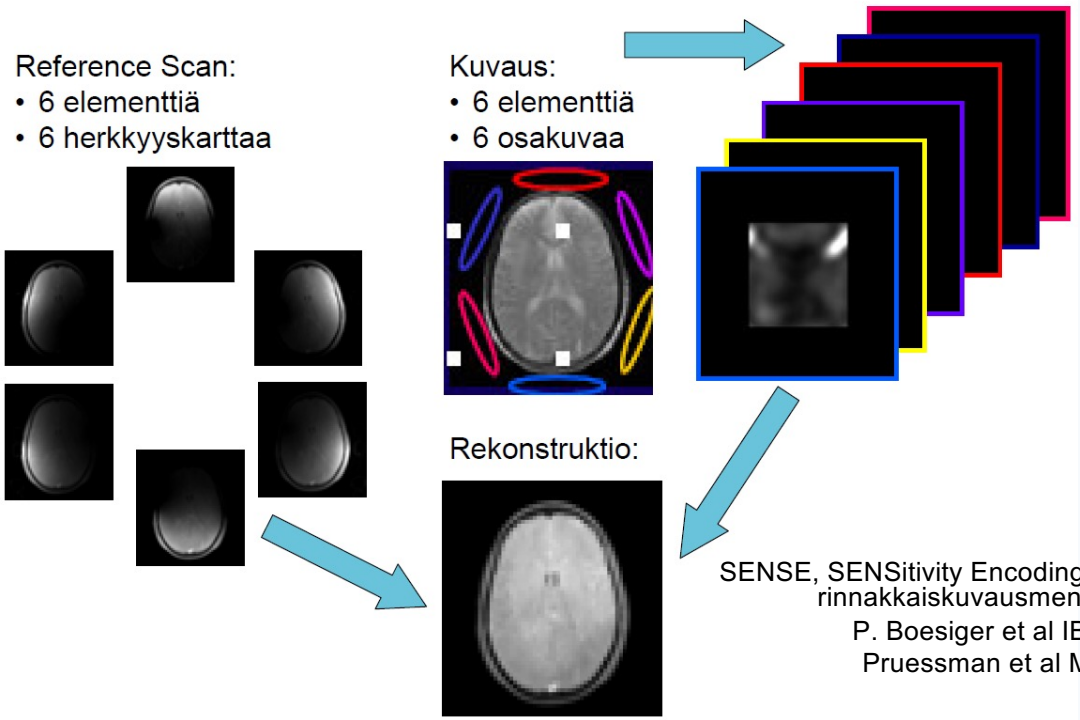
Regular data acquisition vs PROPELLER



Parallel Imaging

18.5.2022

Parallel Imaging

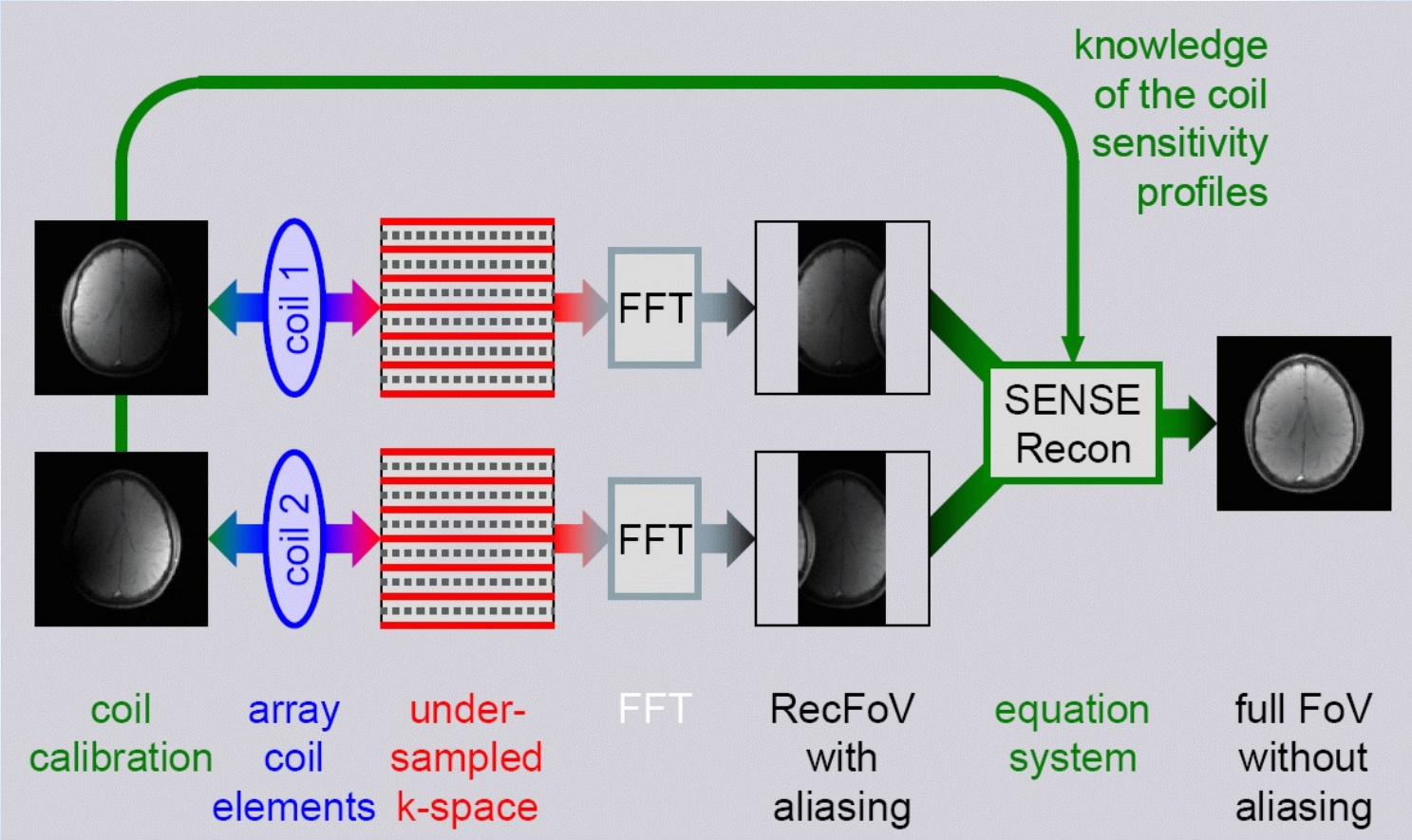


SENSE, SENSitivity Encoding 1. kaupallinen rinnakkaiskuvausmenetelmä

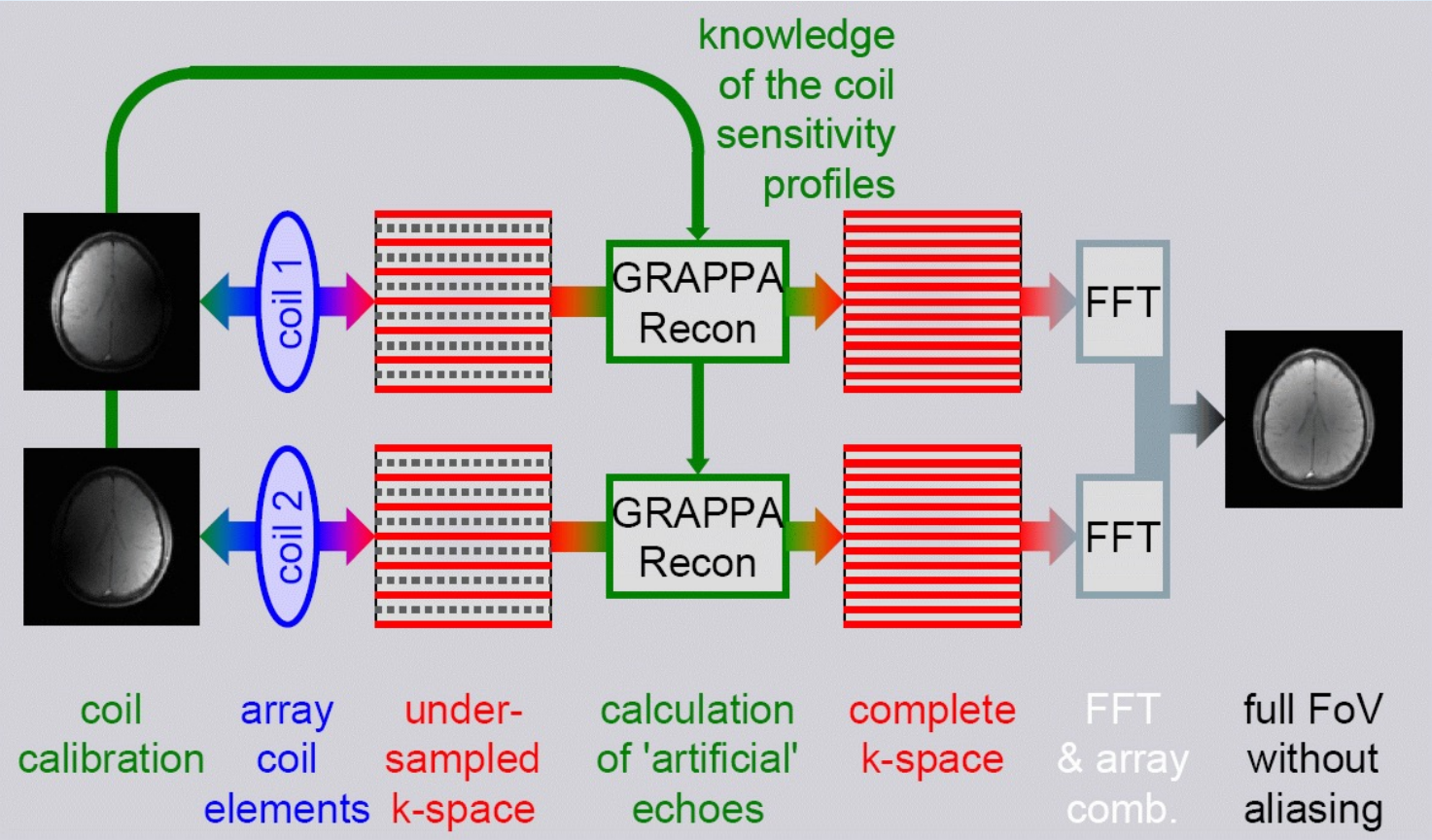
P. Boesiger et al IBTZ, Zurich

Pruessman et al MRM 1999

SENSE

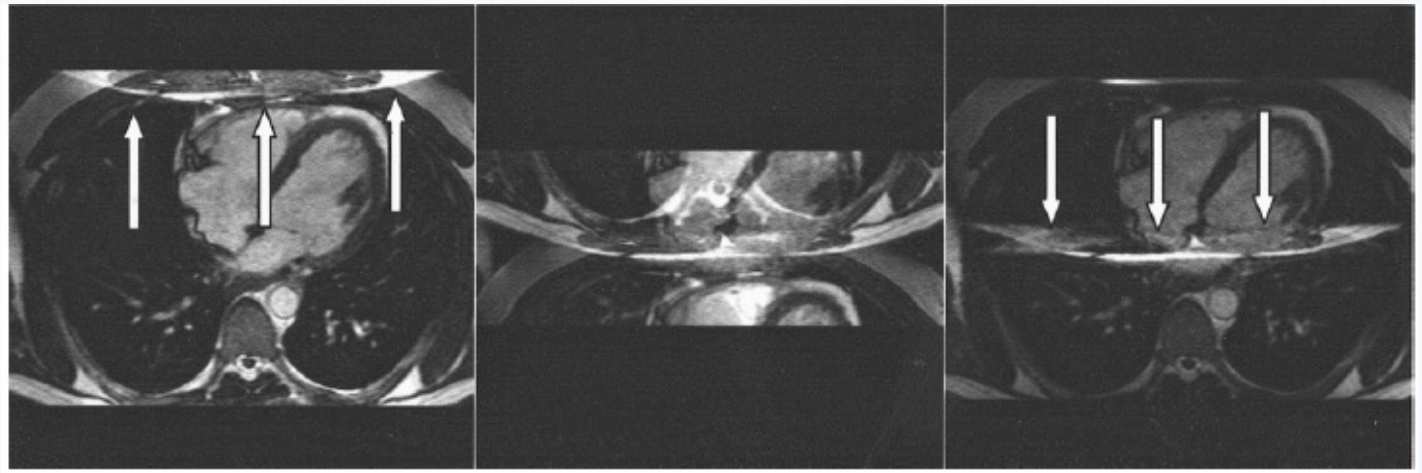

knowledge of the coil sensitivity profiles
coil 1
coil 2
FFT
FFT
SENSE Recon
coil calibration
array coil elements
under-sampled k-space
FFT
RecFoV with aliasing
equation system
full FoV without aliasing

GRAPPA


knowledge of the coil sensitivity profiles
coil 1
coil 2
GRAPPA Recon
GRAPPA Recon
FFT
FFT
coil calibration
array coil elements
under-sampled k-space
calculation of 'artificial' echoes
complete k-space
FFT & array comb.
full FoV without aliasing

SENSE vs GRAPPA

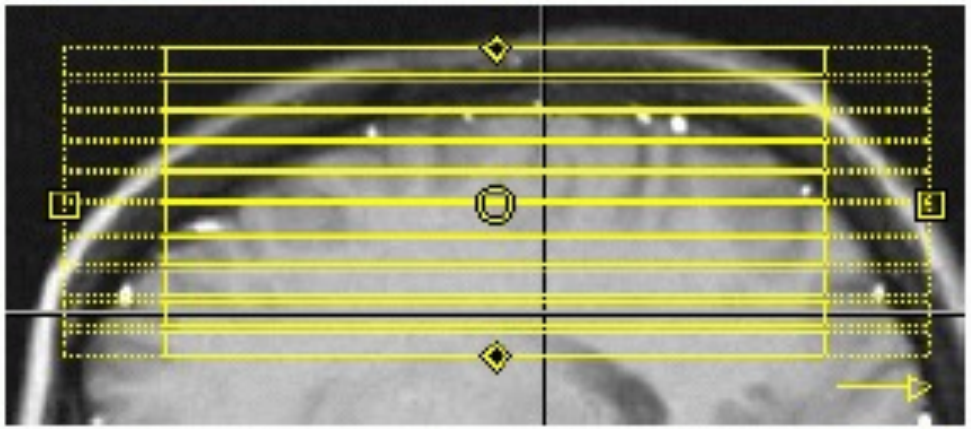
GRAPPA-tekniikassa kelojen herkkyysprofiileista saatu informaatio hyödynnetään siis jo k-avaruuden täyttämisessä
(luodaan uusia viivoja k-avaruuteen)



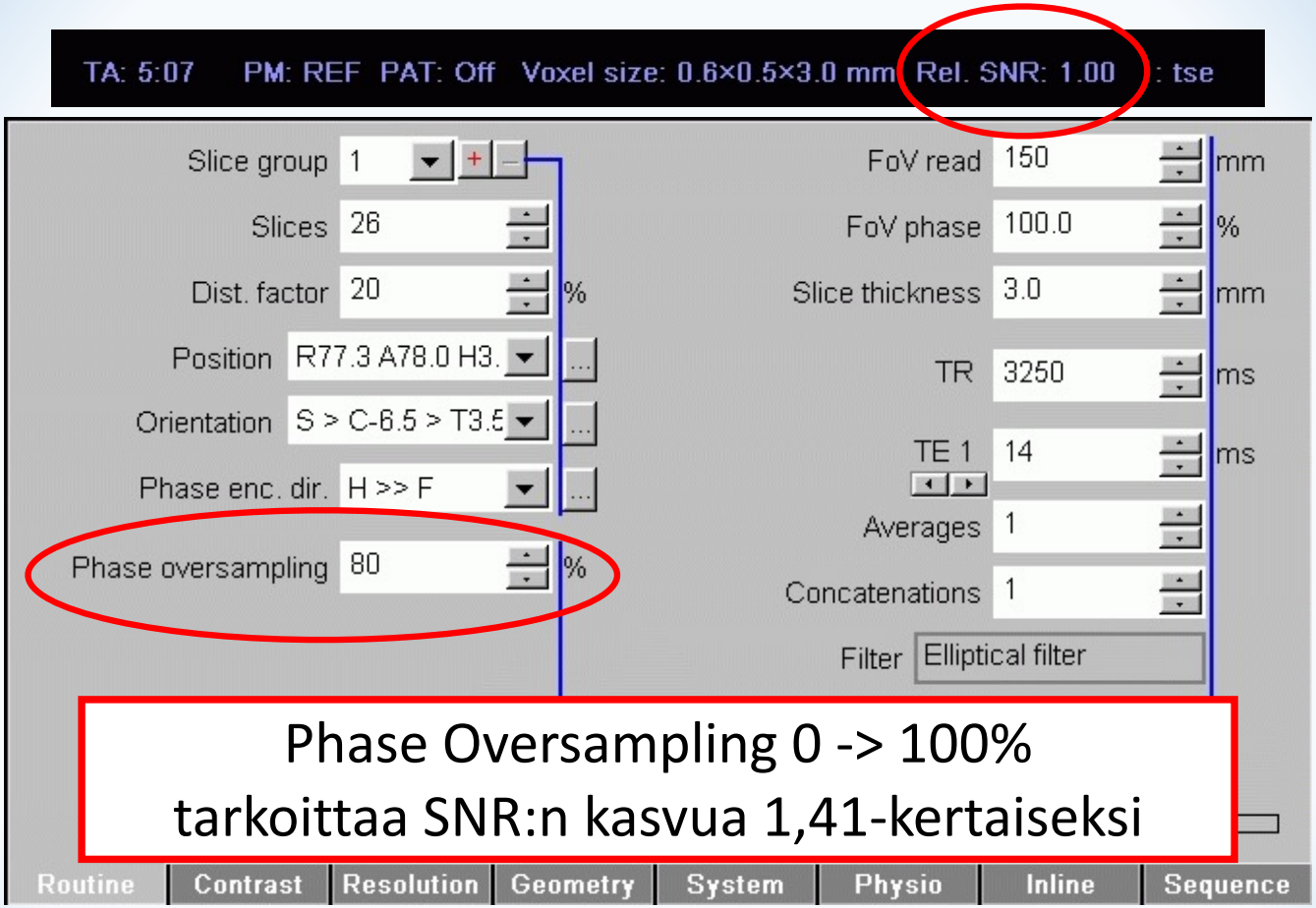
Tällöin vältetään mahdollisesta aliasoitumisesta aiheutuva artefakta, joka SENSE tekniikkaa käytettäessä tulee keskelle kuvaa!

Phase Oversampling / Foldover supression

How to avoid ”aliasing”



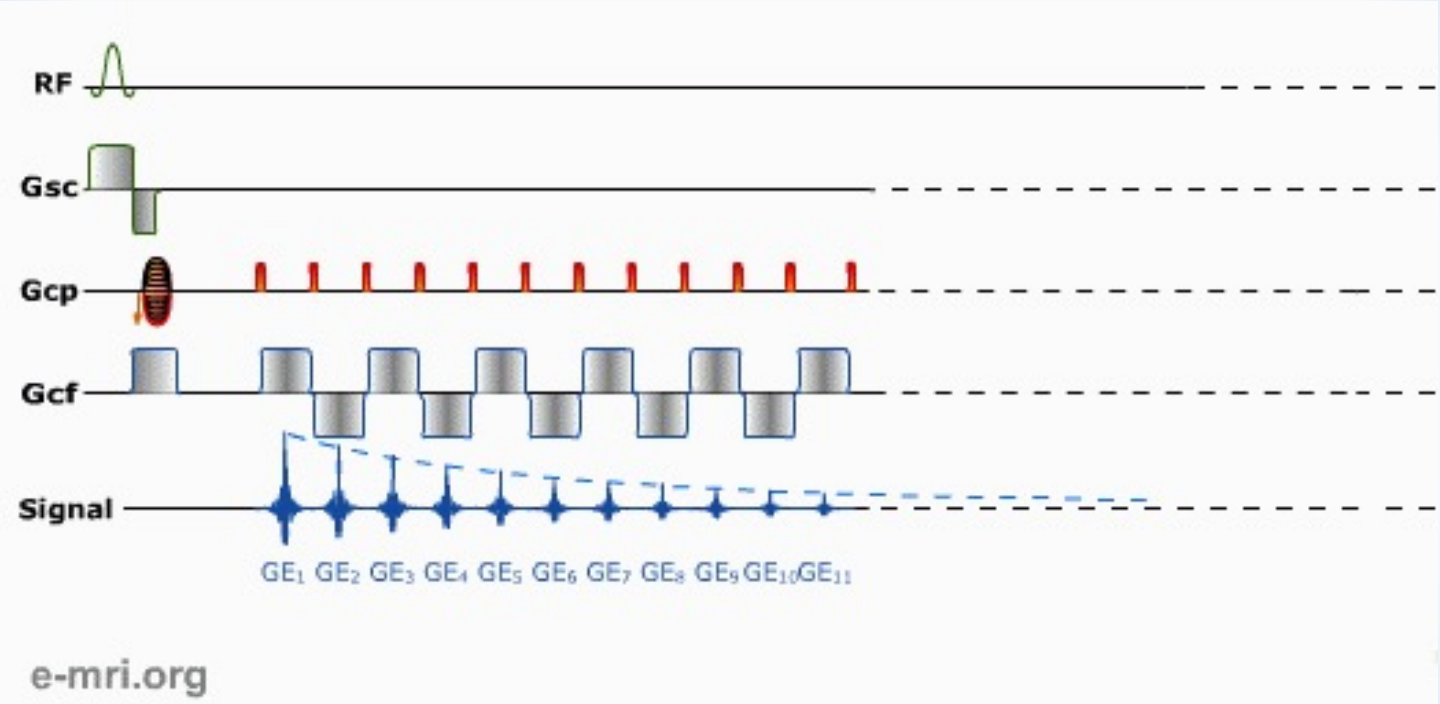
Display of 30% phase-oversampling (2x15%) in the box mode


TA: 5:07 PM: REF PAT: Off Voxel size: 0.6×0.5×3.0 mm Rel. SNR: 1.00 : tse
Slice group 1
Slices 26
Dist. factor 20 %
Position R77.3 A78.0 H3.
Orientation S > C-6.5 > T3.5
Phase enc. dir. H >> F
Phase oversampling 80 %
FoV read 150 mm
FoV phase 100.0 %
Slice thickness 3.0 mm
TR 3250 ms
TE 1 14 ms
Averages 1
Concatenations 1
Filter Elliptical filter
Phase Oversampling 0 -> 100%
tarkoittaa SNR:n kasvua 1,41-kertaiseksi
Routine
Contrast
Resolution
Geometry
System
Physio
Inline
Sequence

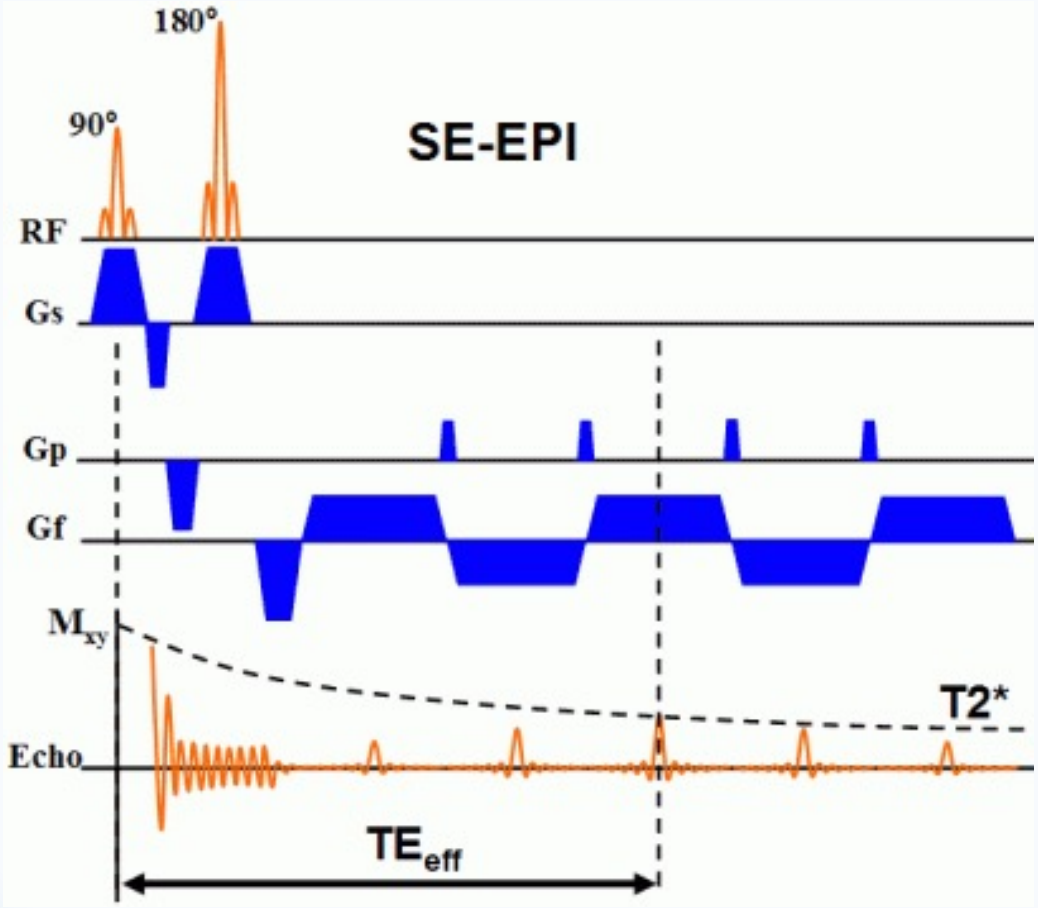
Extremely fast imaging is possible using EPI (Echo Planar Imaging) technique (up to 20 images/second).

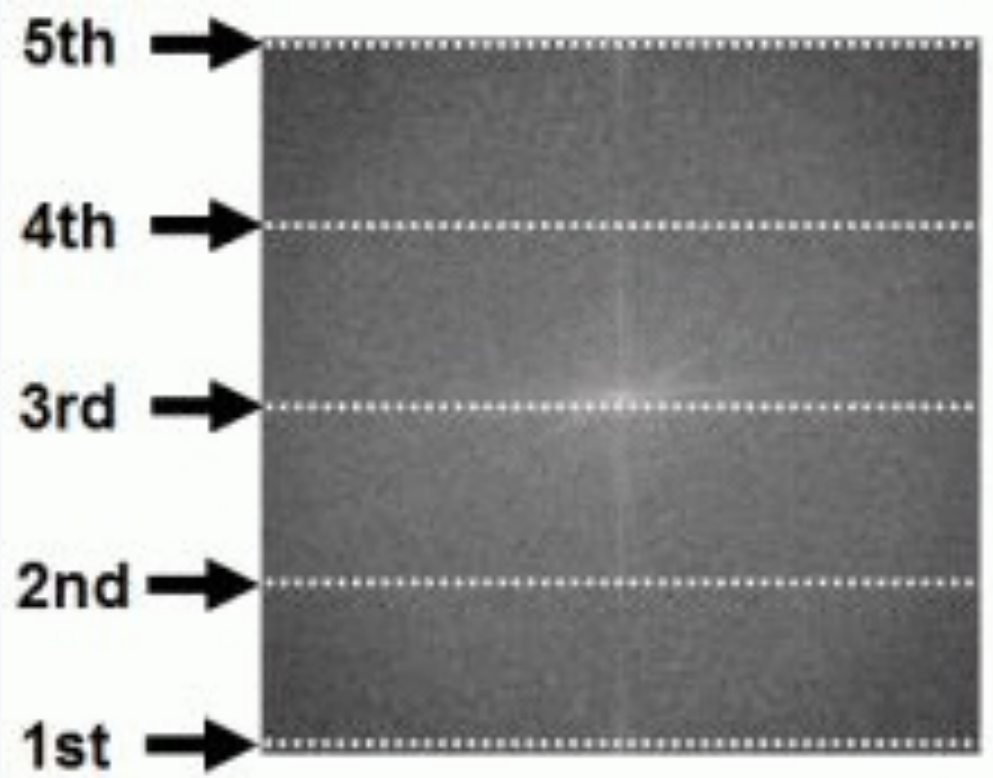
This requires very fast gradients, which can cause neurostimulus to patients.

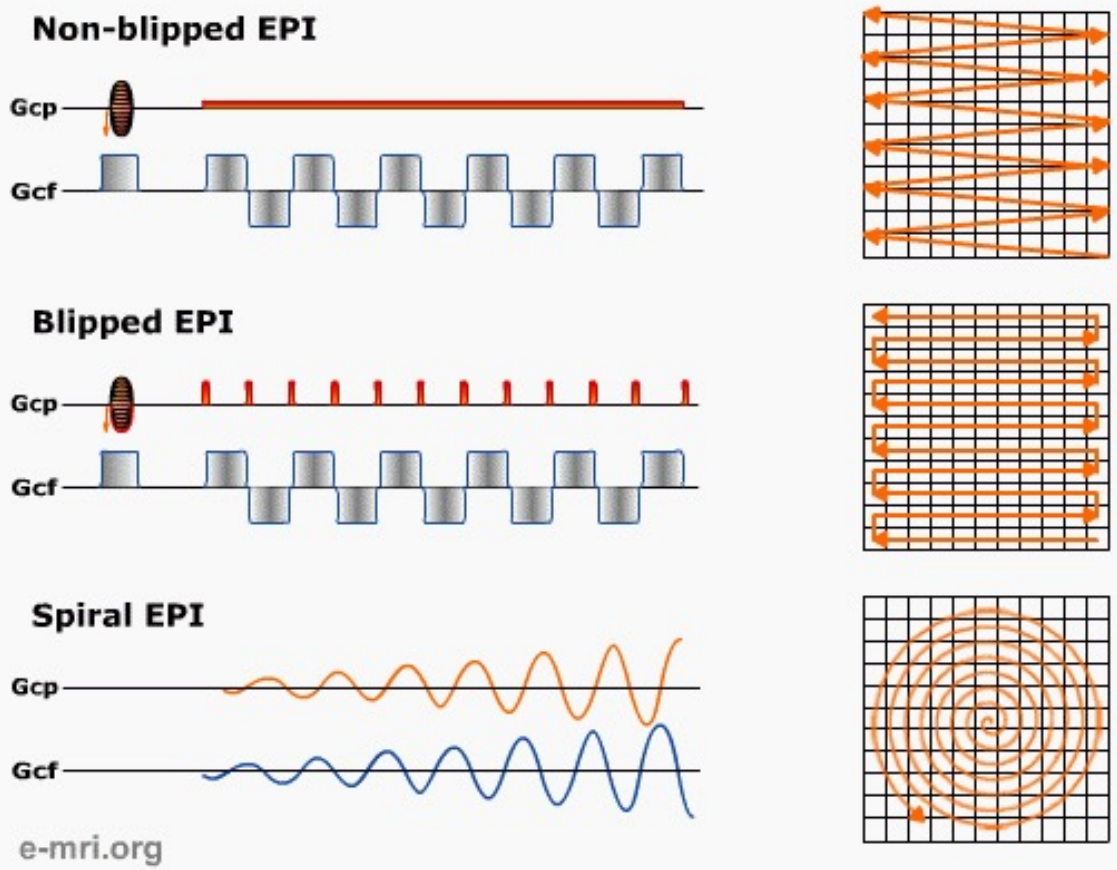
EPI-sequence:



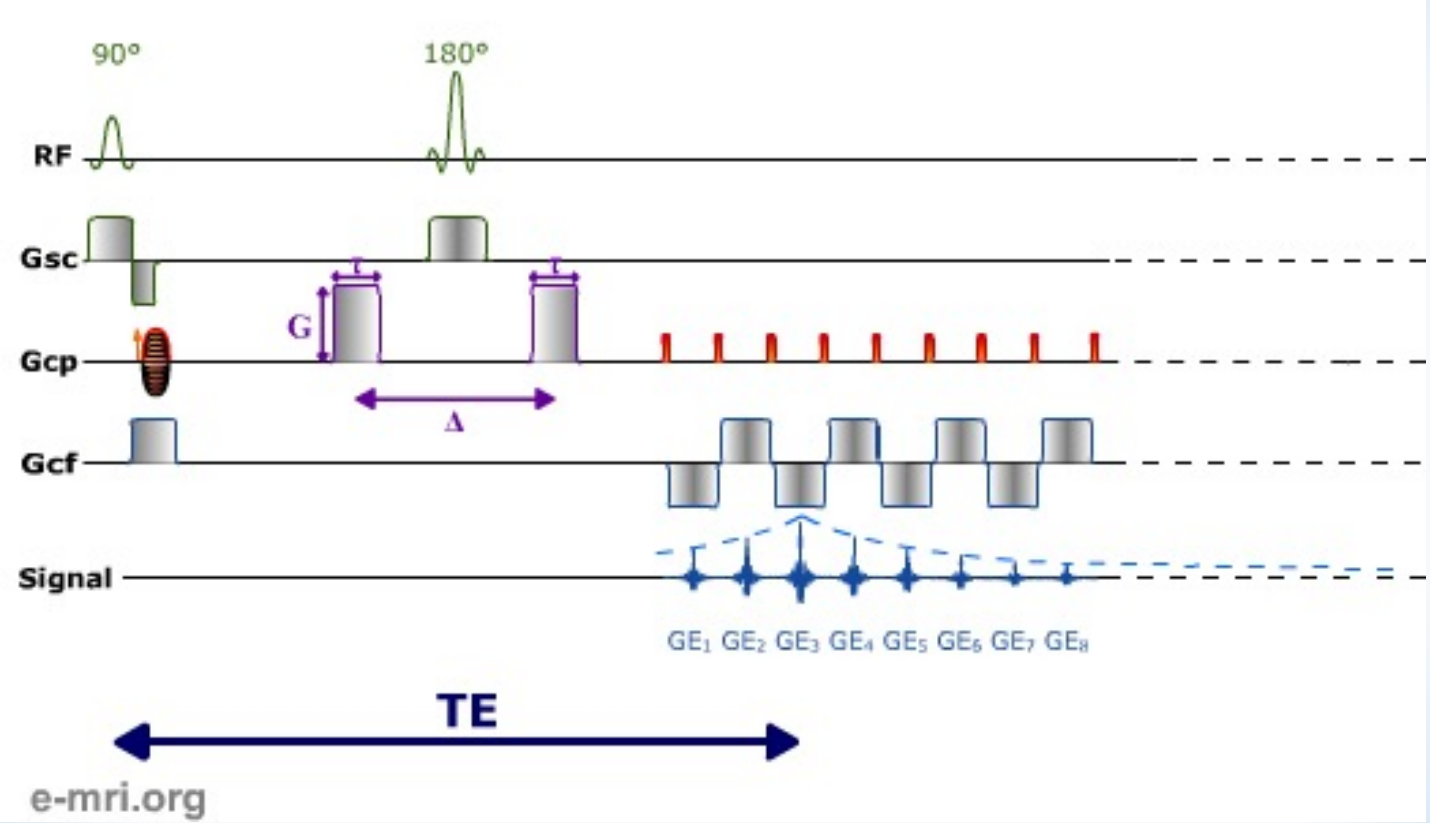
Spin-Echo EPI




5th
4th
3rd
2nd
1st

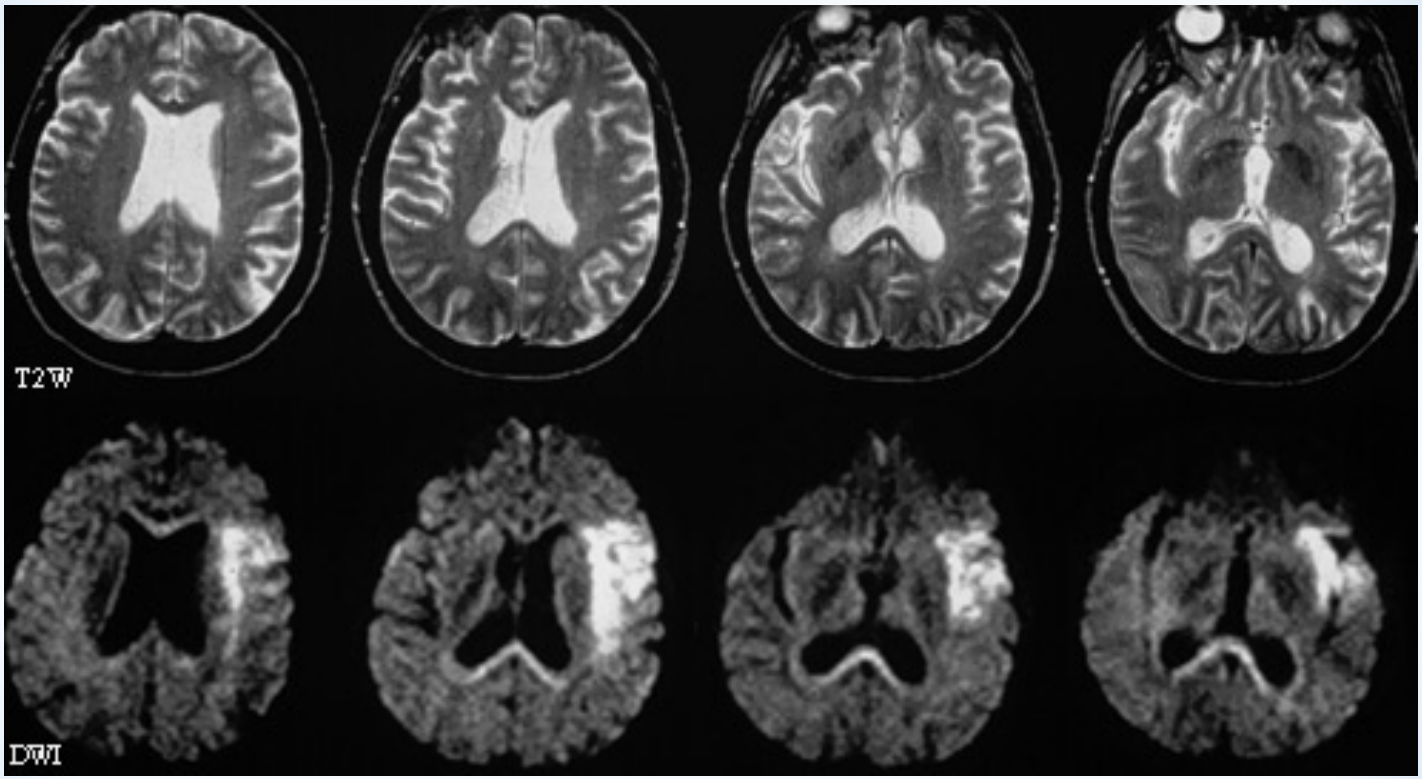

Non-blipped EPI
Gcp
Gcf
Blipped EPI
Gcp
Gcf
Spiral EPI
Gcp
Gcf
e-mri.org

Diffusion Weighted SE-EPI
(additional diffusion encoding gradients)

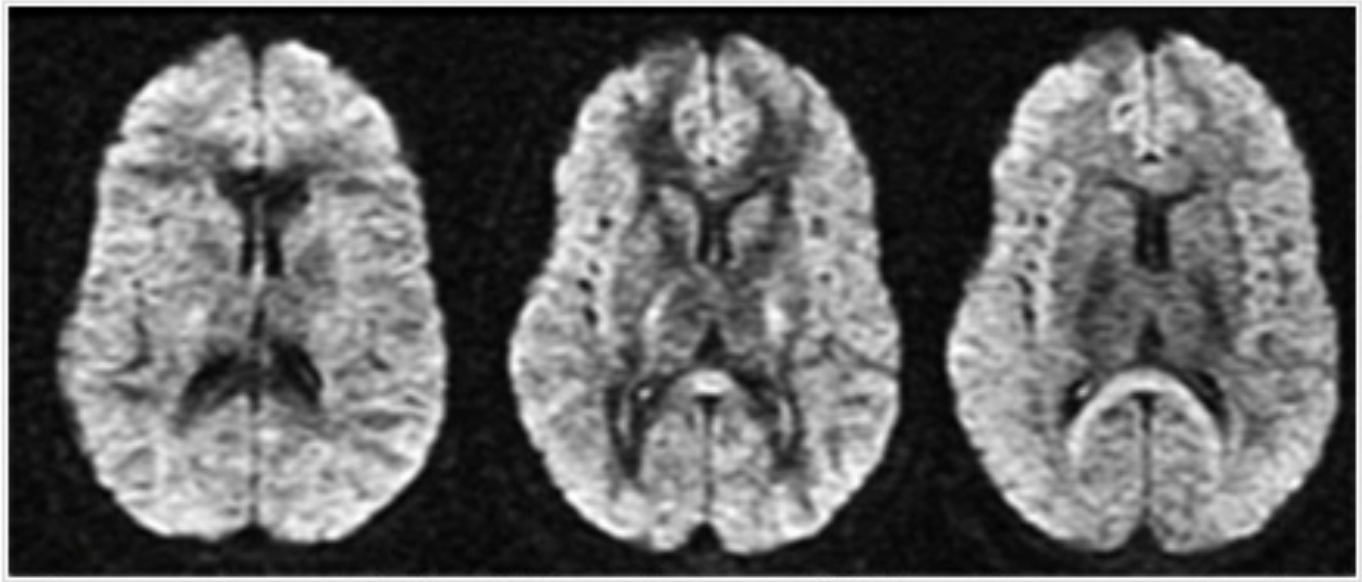


(DWI) Magnetic resonance imaging is sensitive to diffusion, because the diffusion of water molecules along a field gradient reduces the MR signal. In areas of lower diffusion the signal loss is less intense and the display from this areas is brighter. The use of a bipolar gradient pulse and suitable pulse sequences permits the acquisition of diffusion weighted images (images in which areas of rapid proton diffusion can be distinguished from areas with slow diffusion).

Based on echo planar imaging, multislice DWI is today a standard for imaging brain infarction. With enhanced gradients, the whole brain can be scanned within seconds. The degree of diffusion weighting correlates with the strength of the diffusion gradients


T2W
DWI

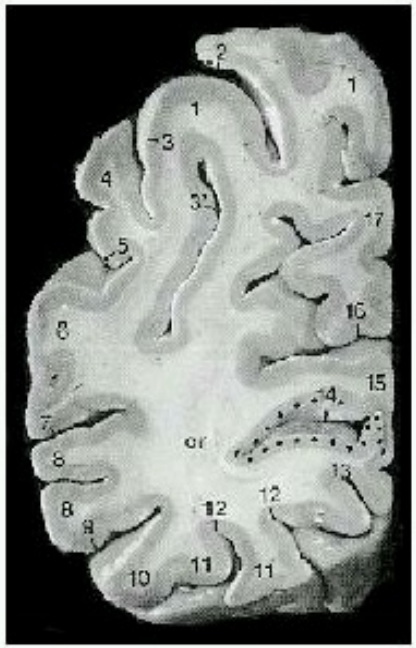
Direction of diffusion encoding gradient



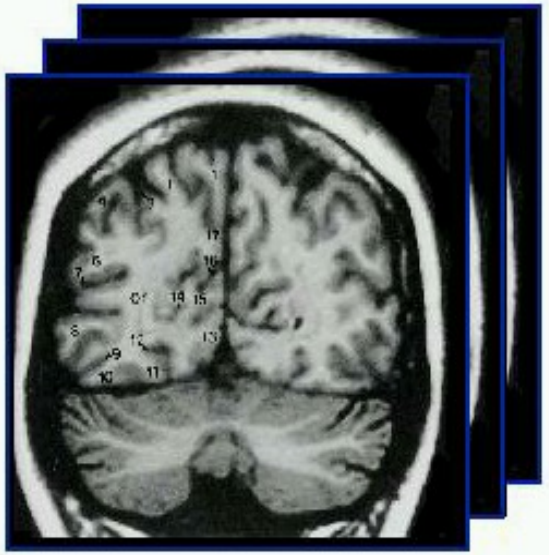
Diffusion source images obtained with gradients applied along the x-, y-, and z-directions

Functional MRI *fMRI*

Structural MRI



Photograph of Cortex



MRI of Cortex

fMRI: BOLD- method

BOLD = **B**lood **O**xygen **D**ependent **L**evel;

- oxyhemoglobin HbO_2 is a diamagnetic molecule;
- deoxyhemoglobin Hb is a paramagnetic molecule;

Cerebral activation <u>decreases</u> paramagnetic [Hb]:

<u>Decrease</u> in transversal relaxation rate R2* = 1/T2* :

$$\Delta R2^* = \left(\frac{1}{T2^*}\right)_{rest} - \left(\frac{1}{T2^*}\right)_{act}$$

$T2^*_{rest}$ = transversal relaxation time T2* in rest phase

$T2^*_{act}$ = transversal relaxation time in activation phase

fMRI: BOLD- method

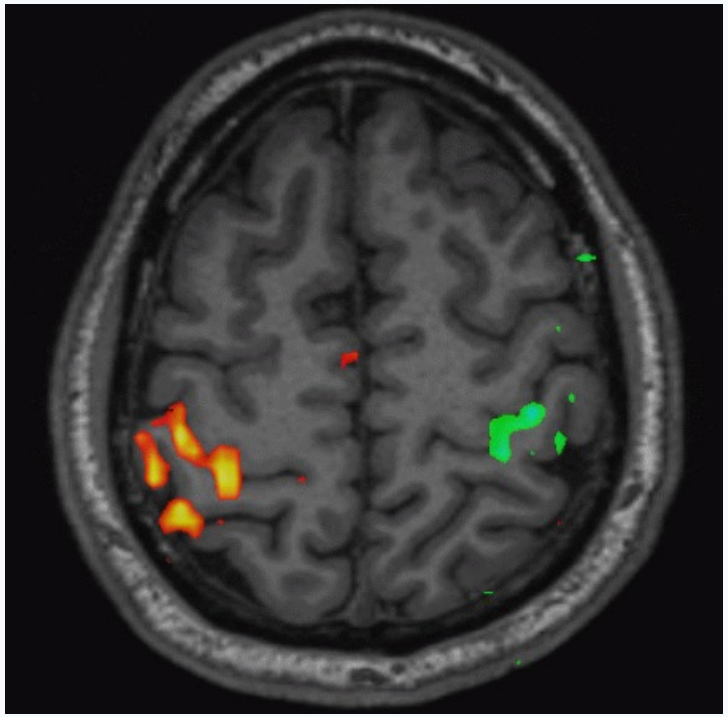
Decrease in transversal relaxation rate R2* = 1/T2* :

$$\Delta R2^* = \left(\frac{1}{T2^*}\right)_{rest} - \left(\frac{1}{T2^*}\right)_{act}$$

Increase in T2*-time increase in gradient echo GRE MR image intensity in the activated area;

Enhanced image intensity during neural activation phase;

Statistical methods used to compare the pixel intensities during rest and activation phases fMR image



Applications

- Display of neurally active regions of the brain due to motor or sensory stimulation and/or cognitive processes
- Analysis of the cerebral organization of functional systems
- Examination of changes in activity due to local brain lesions
- Identification of functionally important areas prior to brain surgery