MRI basics

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Recommendations for more information

www.imaios.com

www.mriquestions.com

www.revisemri.com

Homework for MRI lectures on 4.6.

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.

AAPM/RSNA Physics Tutorial for Residents

MR Imaging: Brief Overview and Emerging Applications[1]

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

TEACHING POINTS
See last page

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, $T1$
transverse relaxation time, $T2$

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
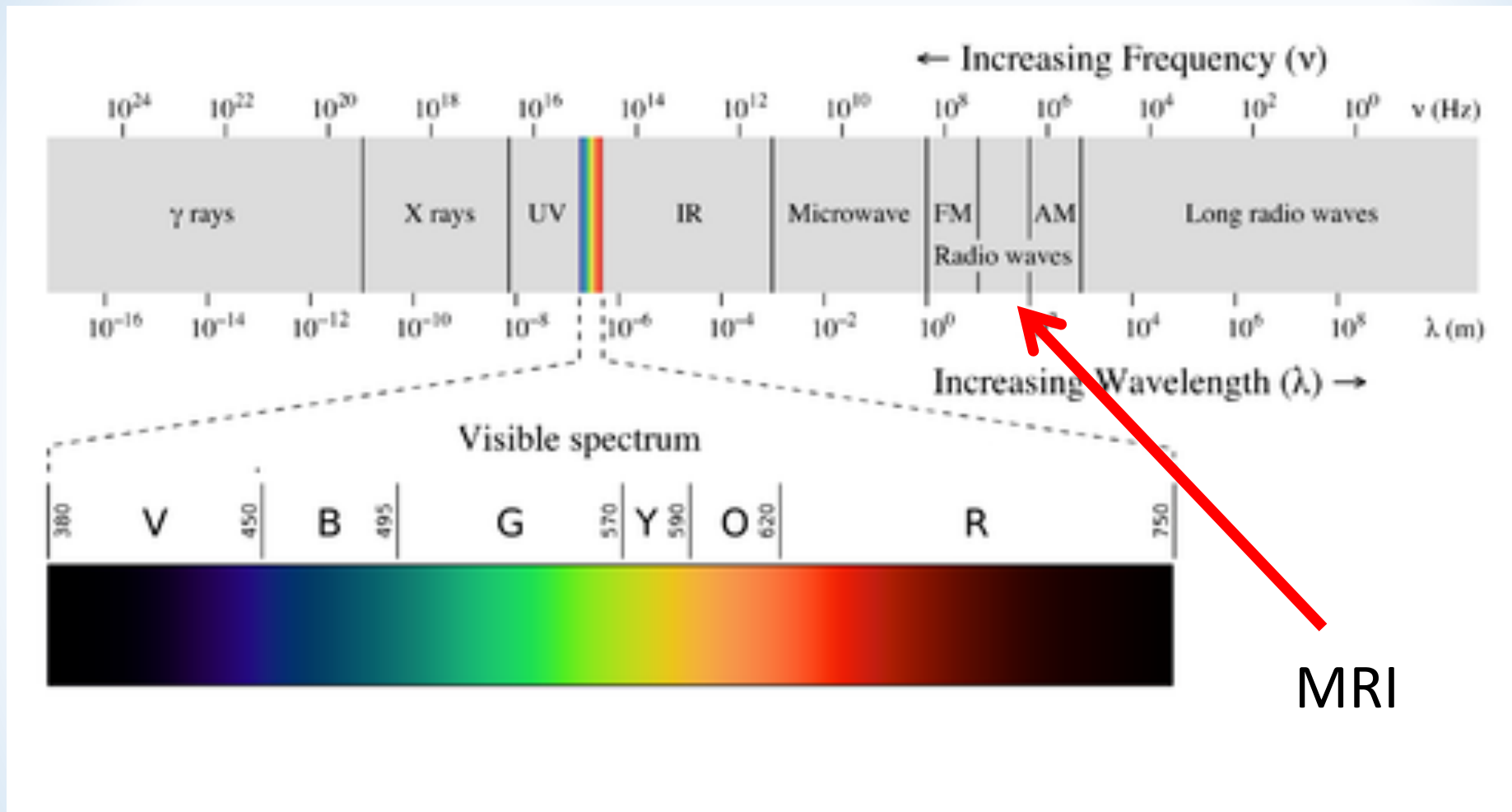
Nuclear Magnetic Resonance

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**.
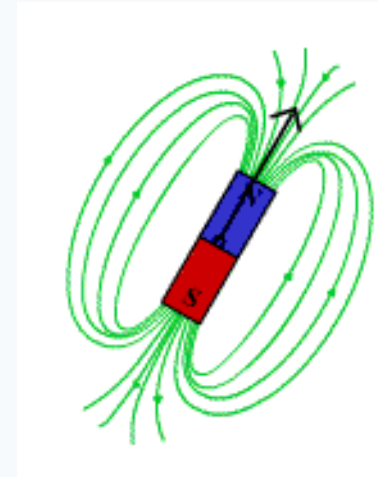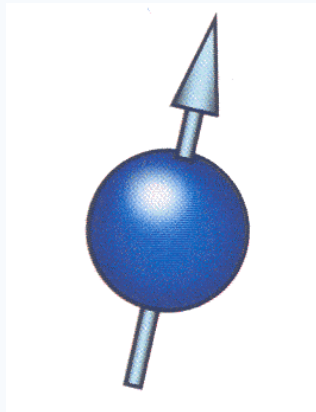
Compass needle in magnetic field
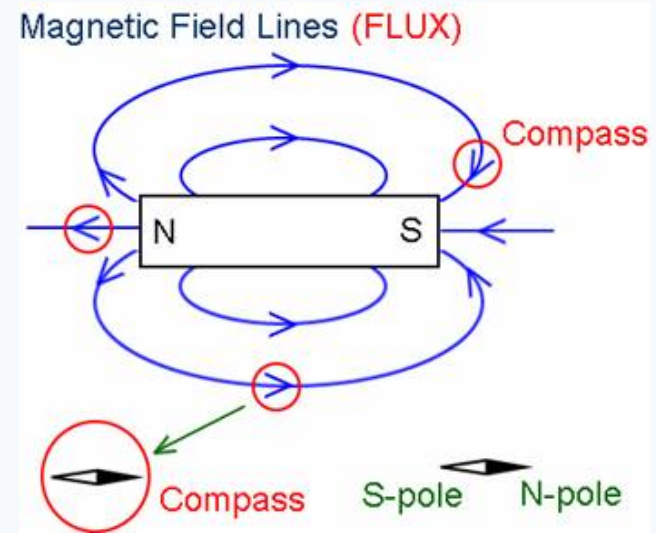
Macroscopic magnetization M$_z$
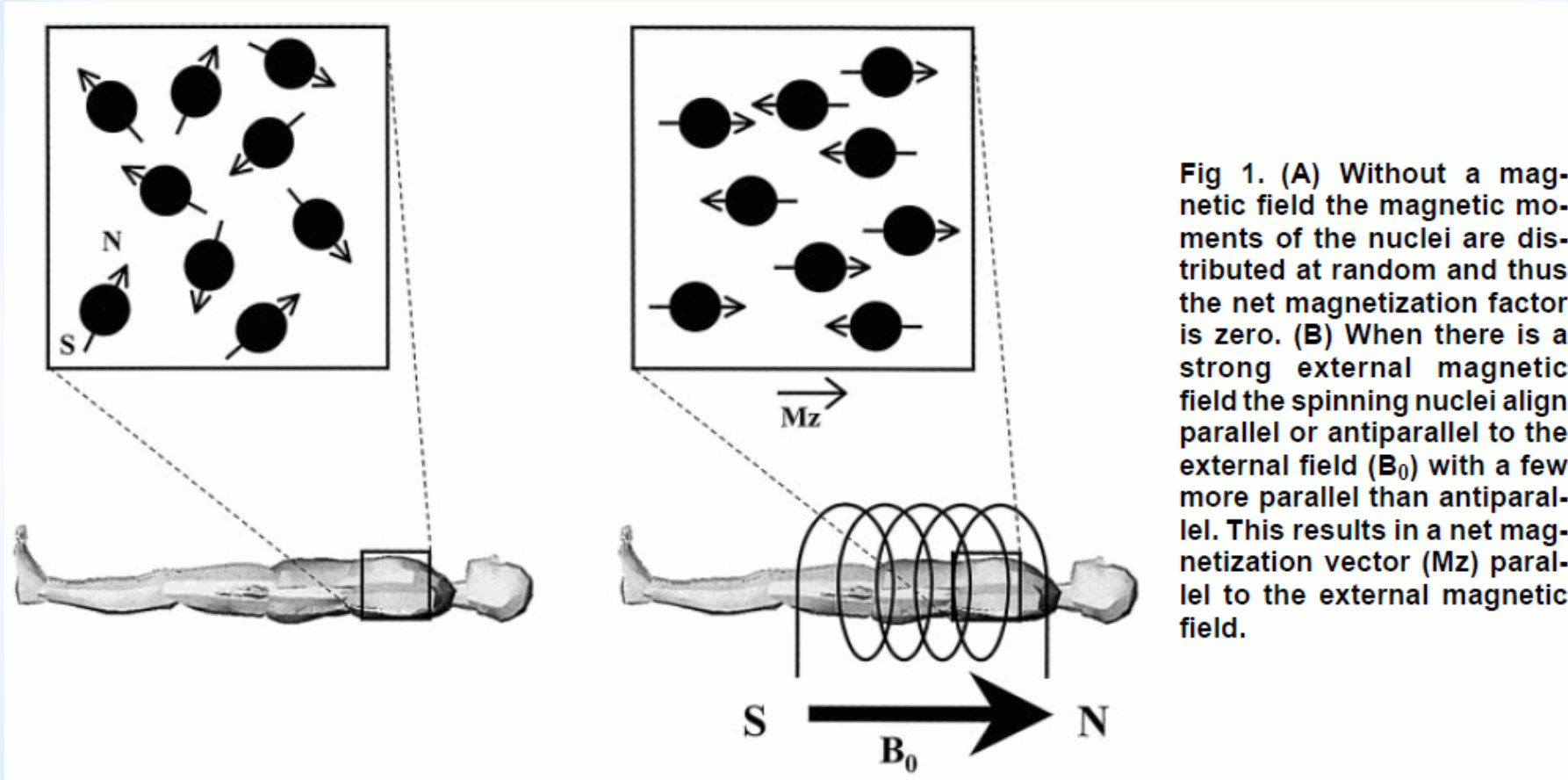
Fig 1. (A) Without a magnetic field the magnetic moments of the nuclei are distributed at random and thus the net magnetization factor is zero. (B) When there is a strong external magnetic field the spinning nuclei align parallel or antiparallel to the external field (B_0) with a few more parallel than 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**.
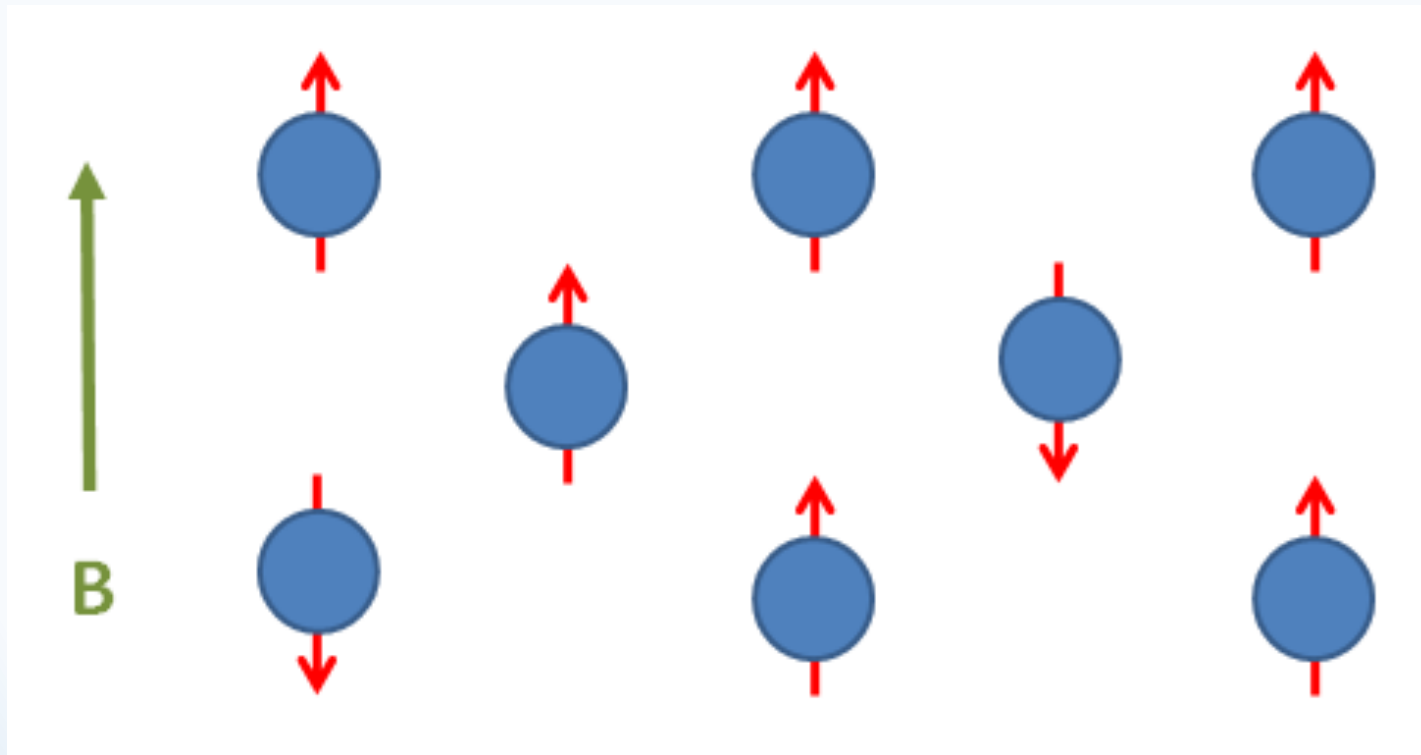
Spin and magnetic field
(quantum mechanical model)

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 ^1H spins in magnetic field
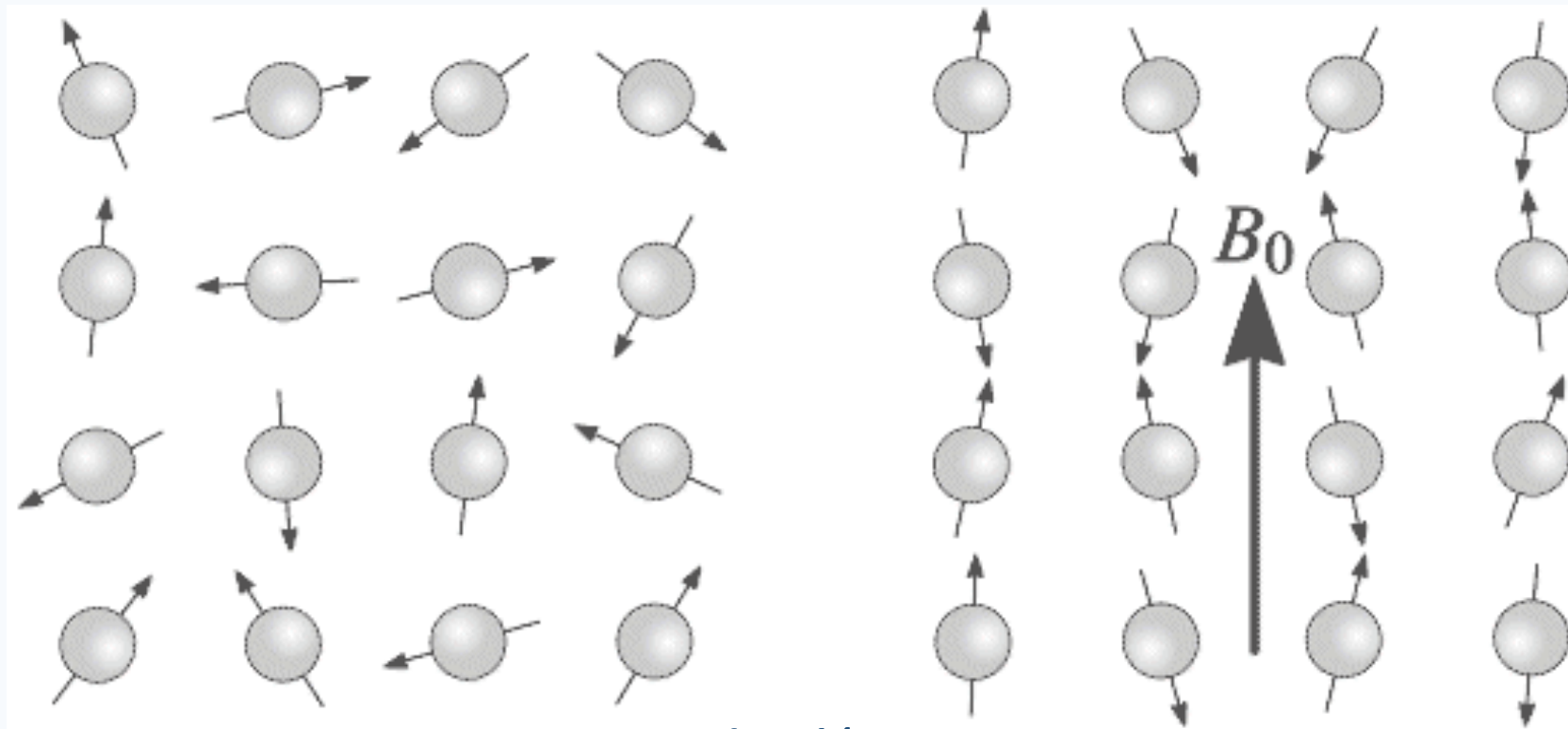
$$I = ½$$
$$\rightarrow 2* ½ + 1 = 2 \text{ sublevels}$$

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)
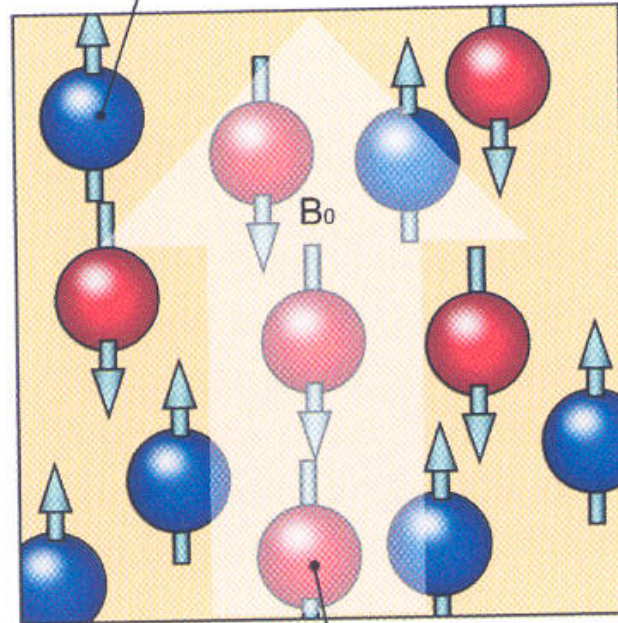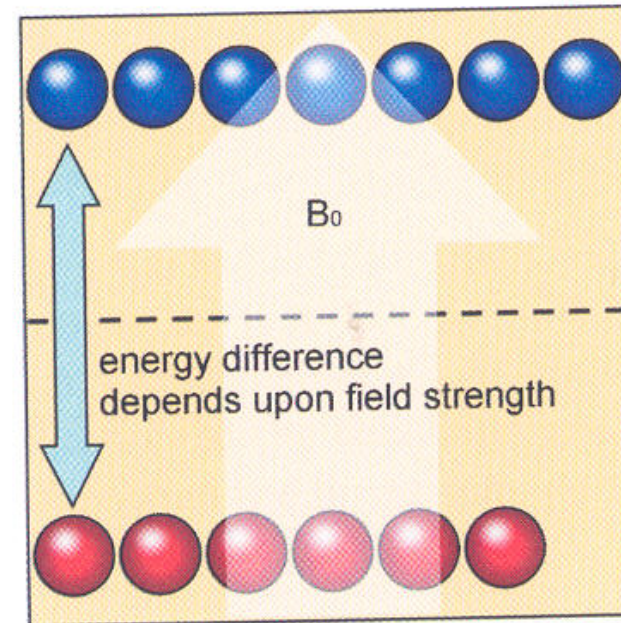
Spin and magnetic field (quantum mechanical model)
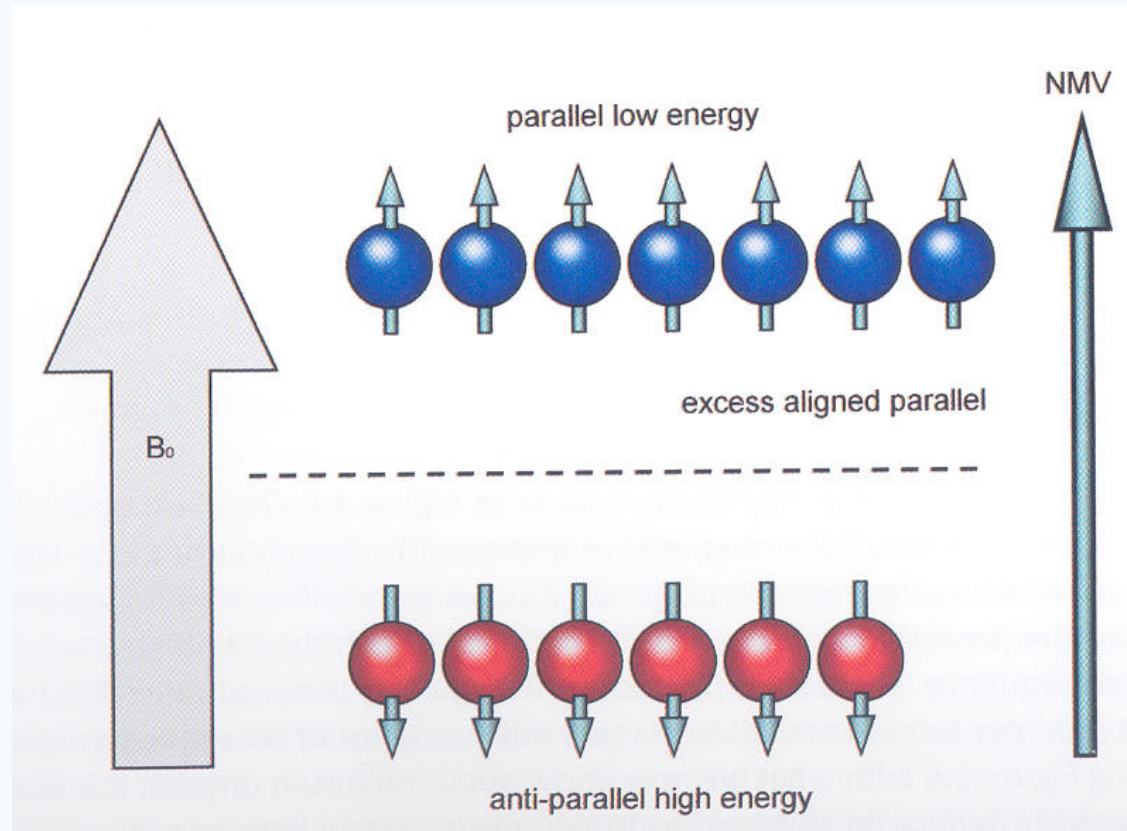
According to **Boltzmann statistics**, the low energy level will have a higher population than the higher energy level, so that **thermal equilibrium** is reached

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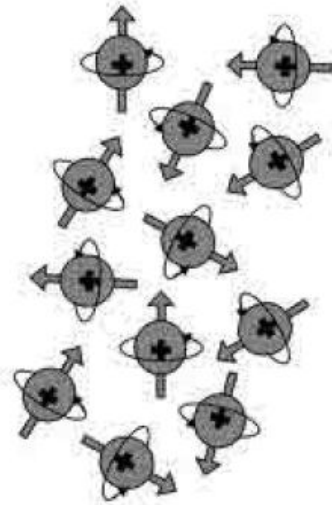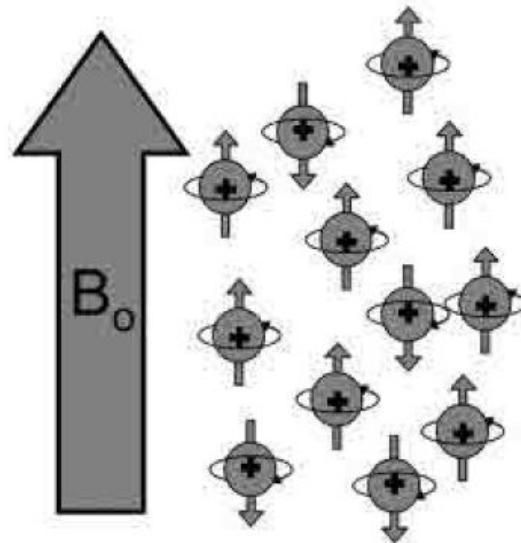
Figure 4. Alignment of protons with the B_0 field. With no external magnetic field, hydrogen protons $(+)$ are oriented randomly. When the protons are placed in a strong magnetic field (B_0), a net magnetization will be produced parallel to the main magnetic field.

Precession motion of the magnetic moment of nucleus in magnetic field

precession

precessional path

B_0

magnetic moment of the nucleus

spinning hydrogen nucleus

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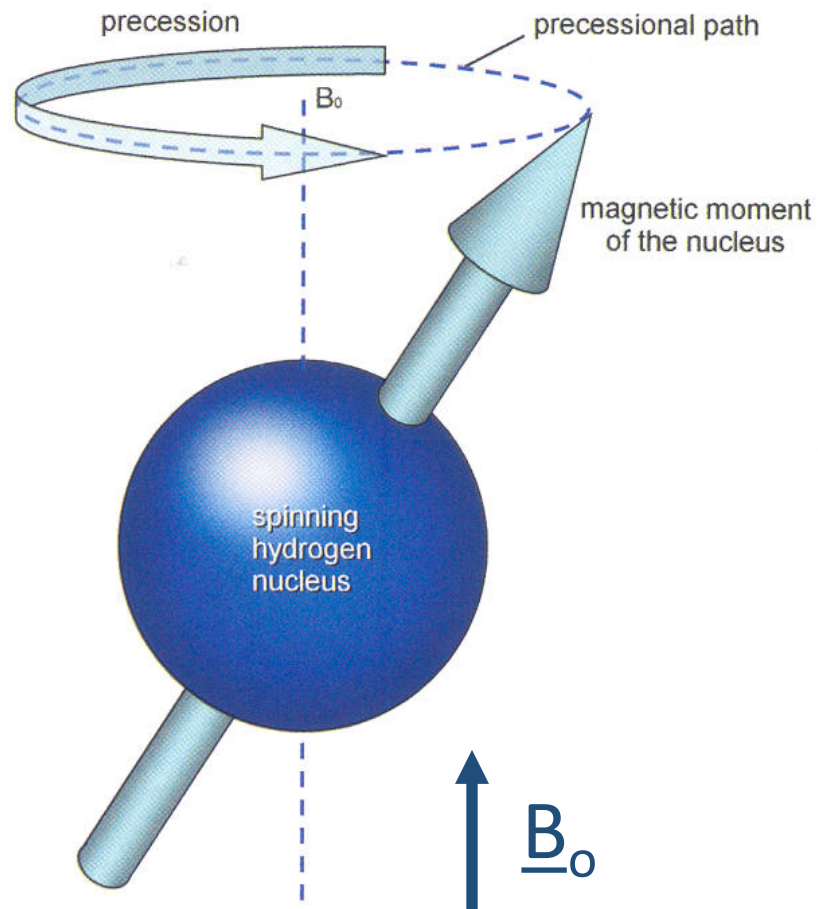
Z-axis is always considered to be in the direction of $\underline{\mathbf{B}}_o$-field

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

ω = Larmor frequency

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

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

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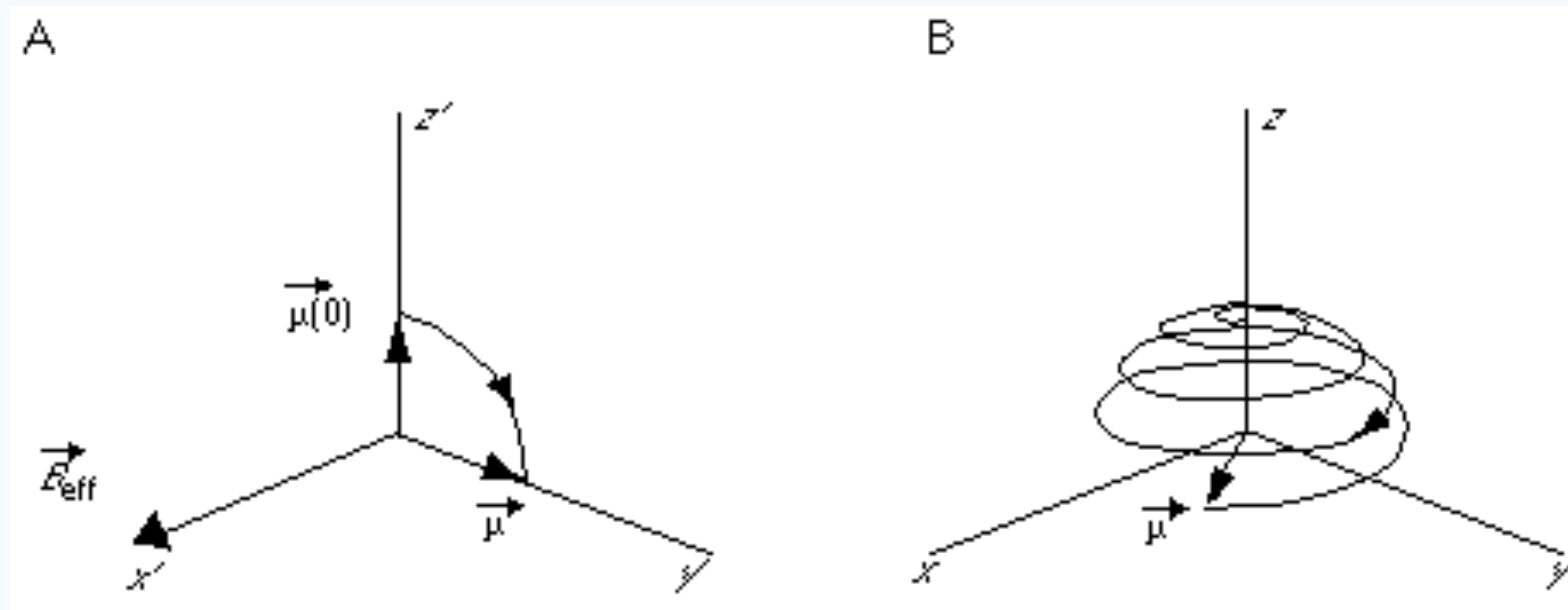
Effect of oscillating (Larmor frequency) RF-field on *M*
(rotating frame of reference)

Geometry of the Fields

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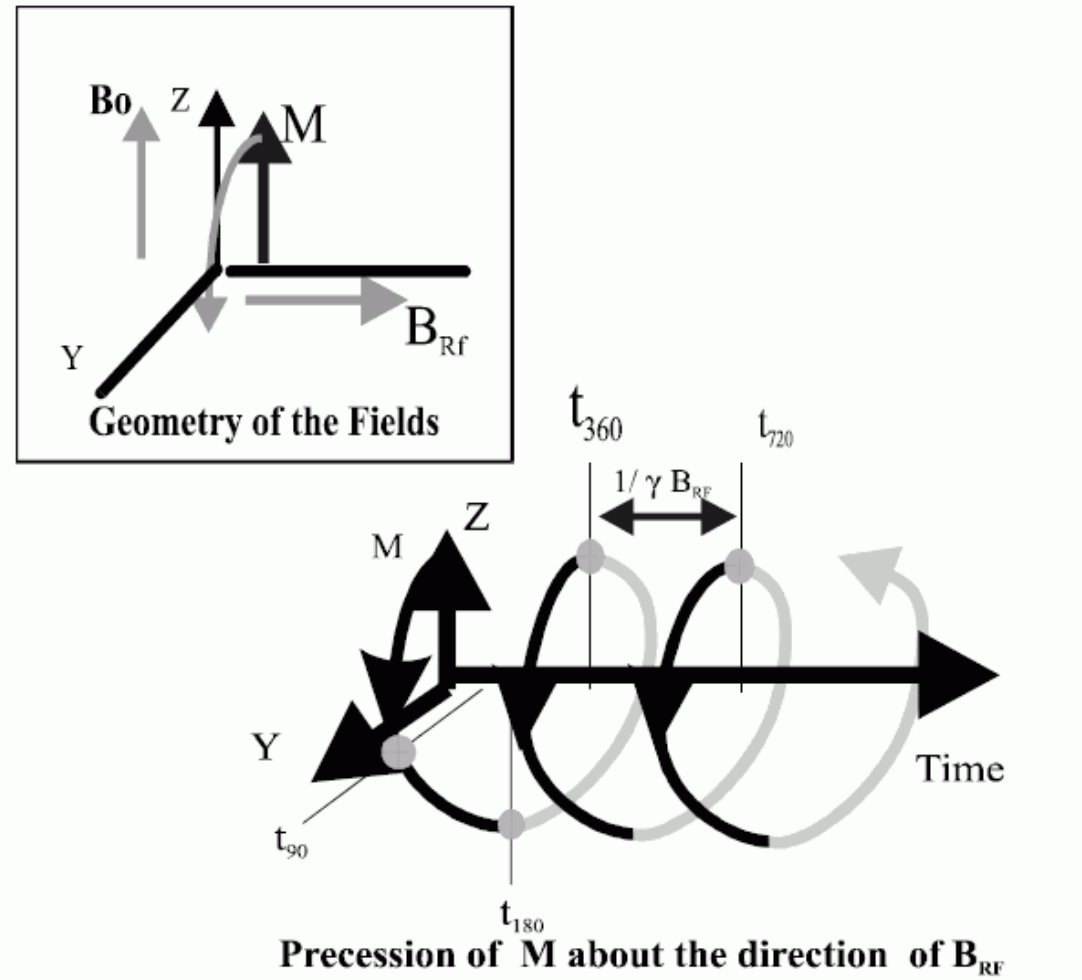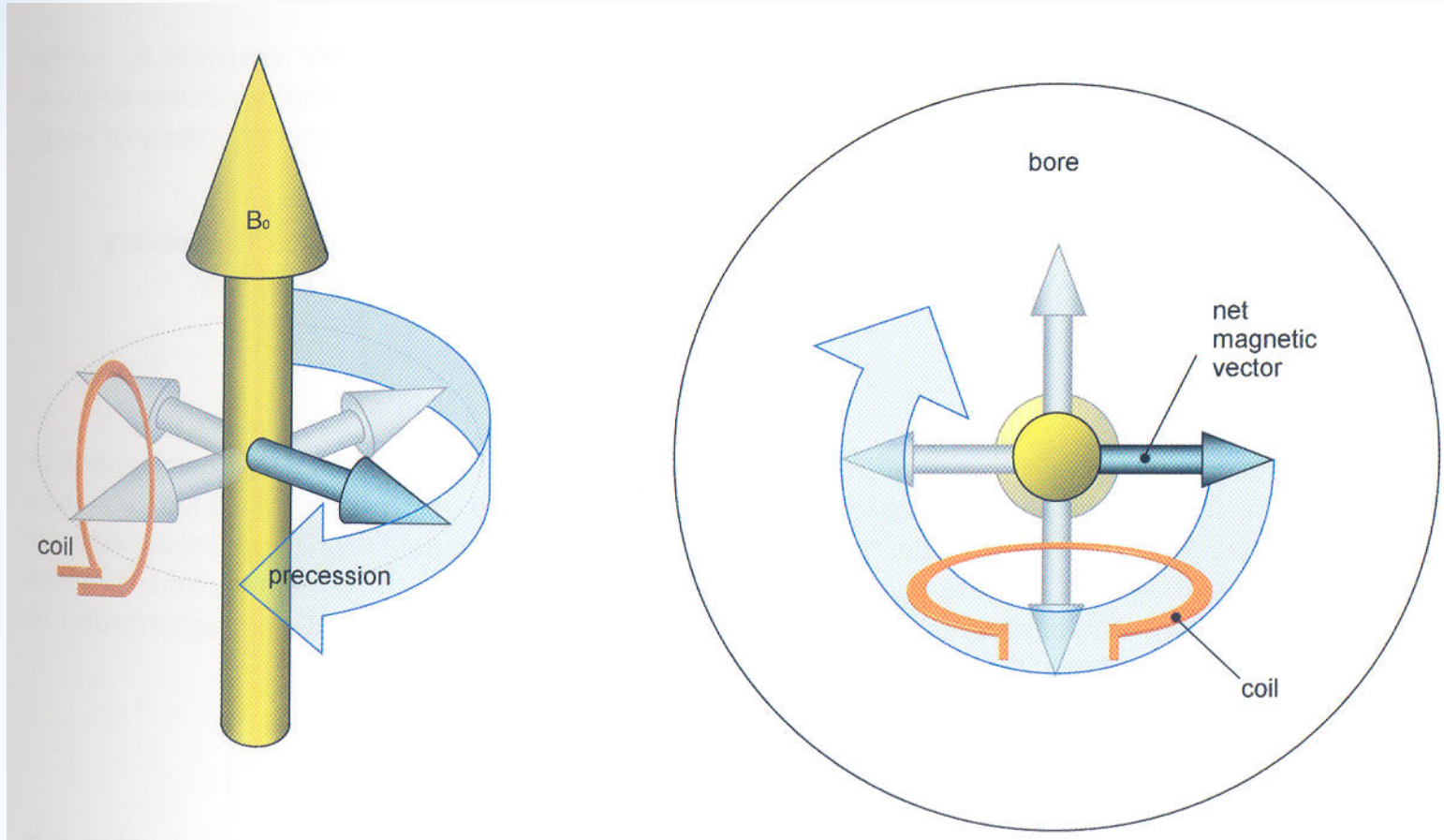
Immediately after 90 degree RF pulse all magnetic moments (spins) are in same phase, so Net Magnetization vector is completely in xy-plane.
Magnetic moments continue their precession motion and if there is a coil against xy-plane, there will be fluctuating magnetic field inside of it...

Electromagnetic induction

Faradays Law of Induction
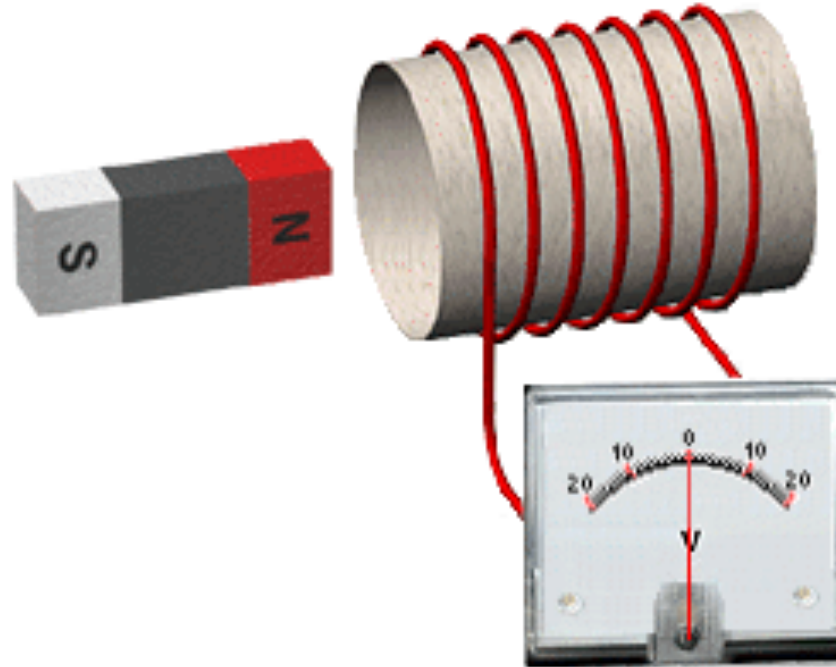
Kieran Mckenzie

Receiver coils

Electromagnetic induction
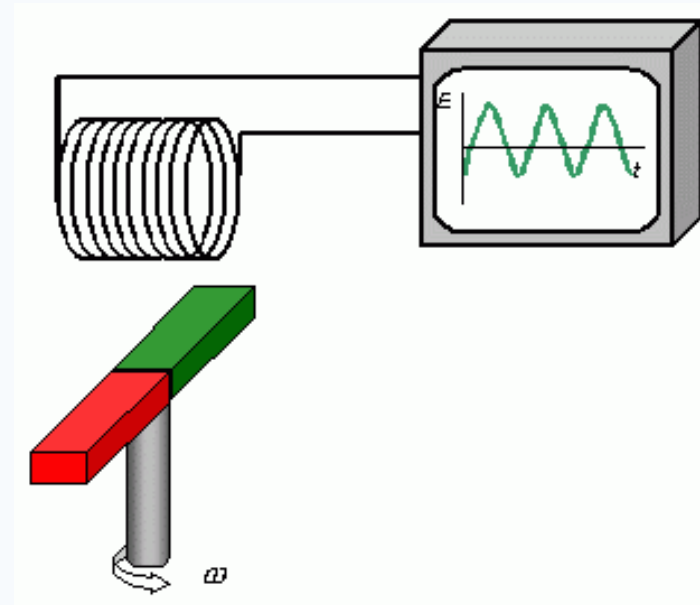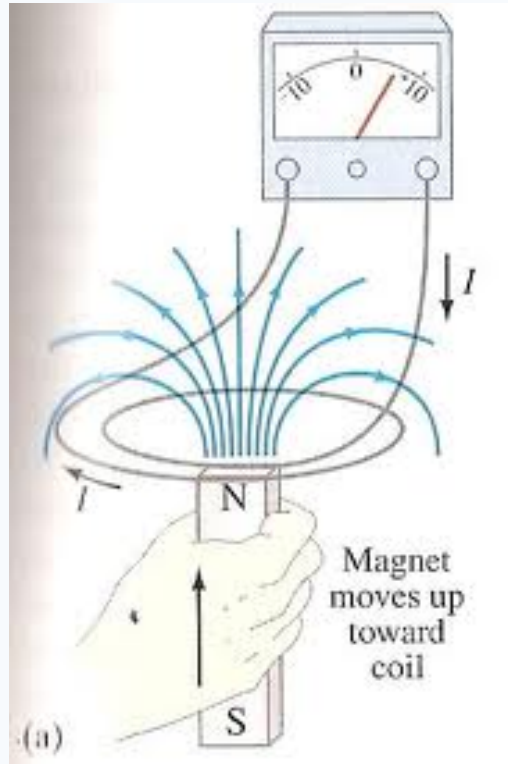
Free Induction Decay -signal
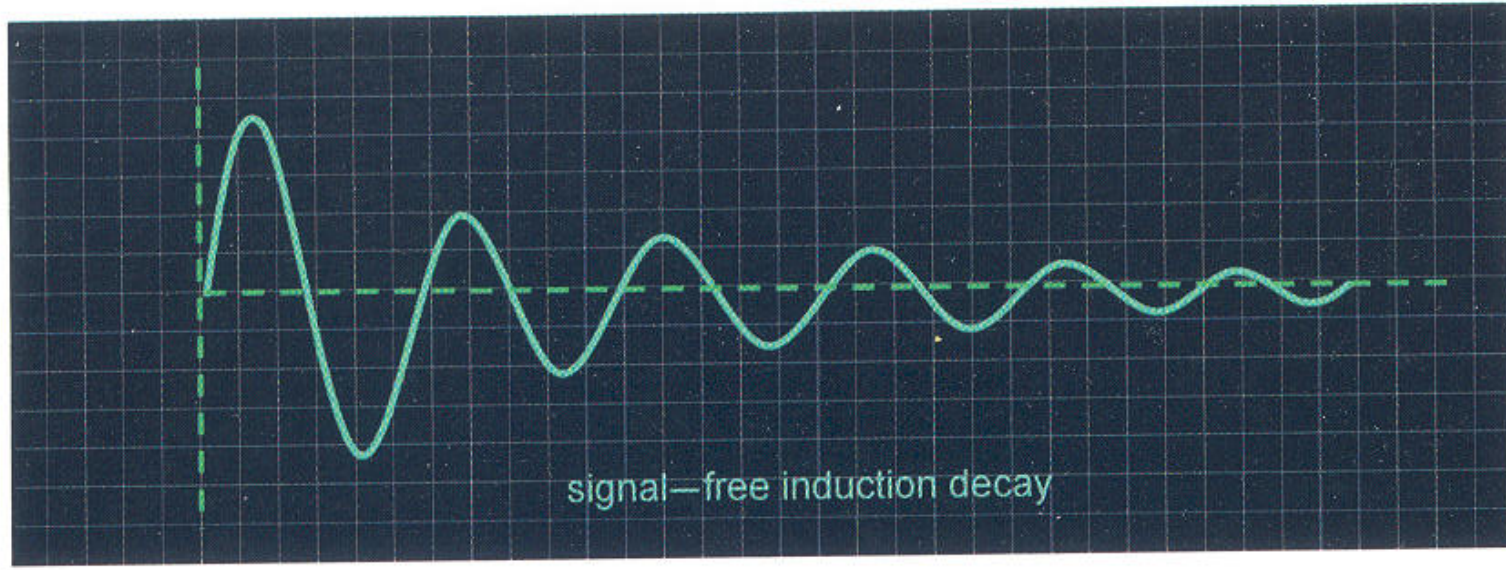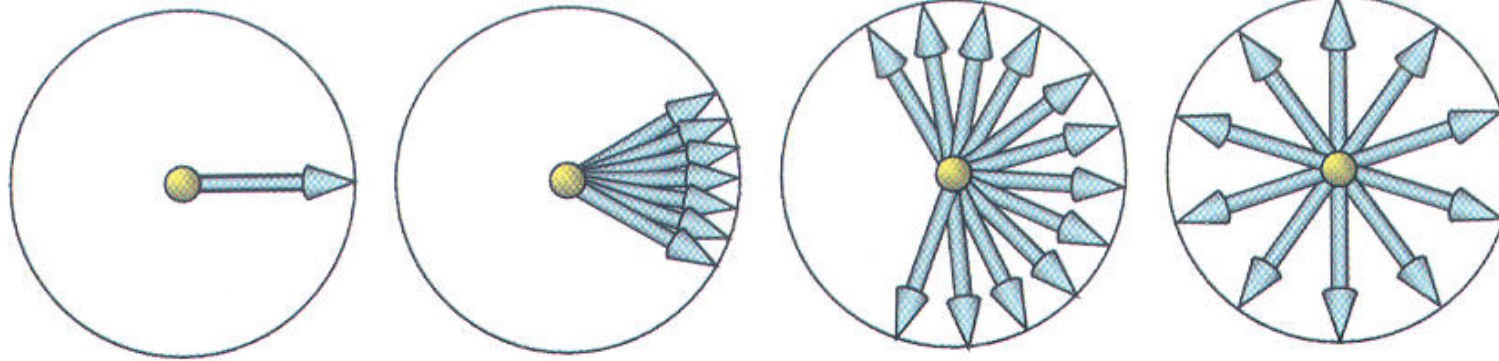
T2* relaxation

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).
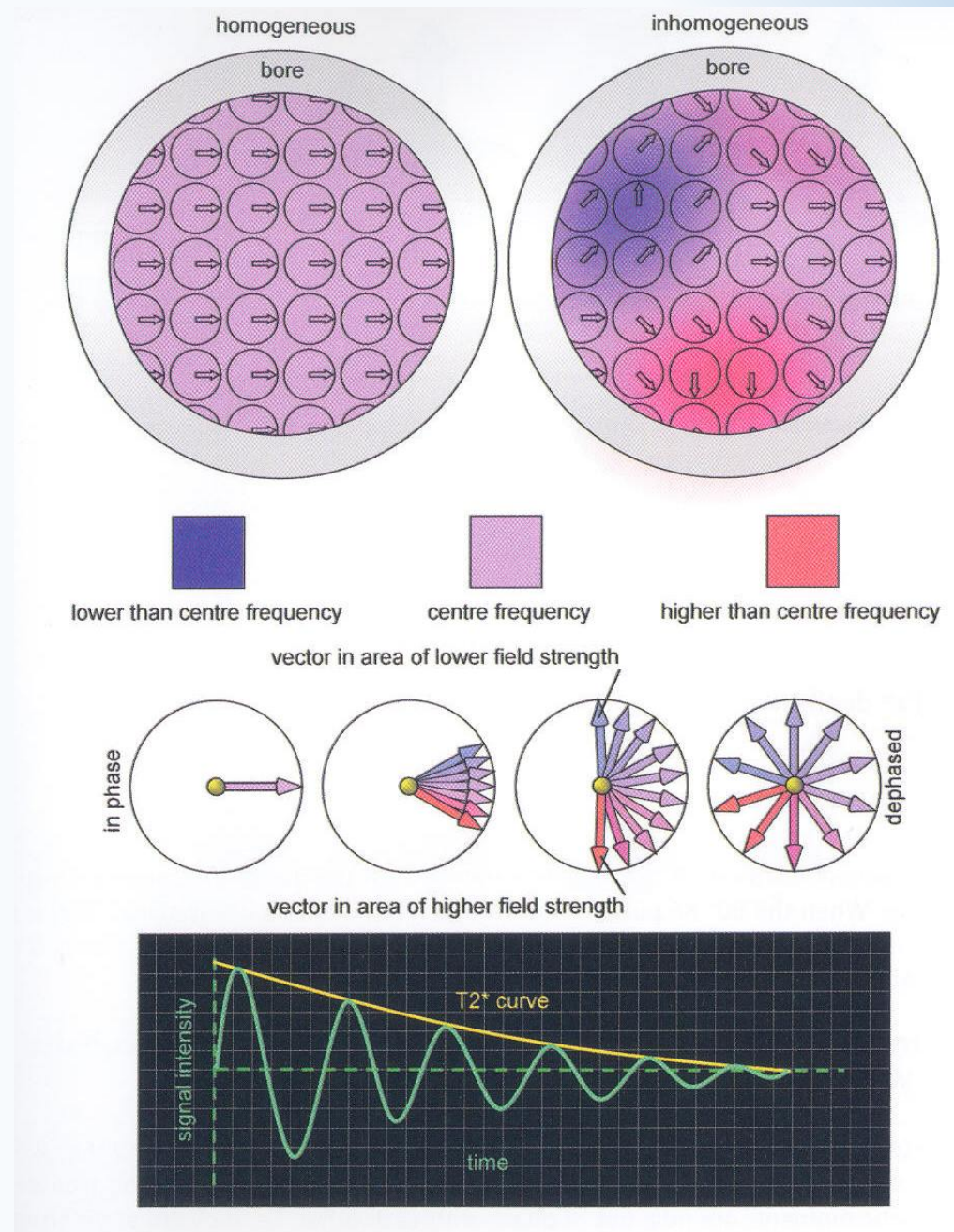
T2* relaxation
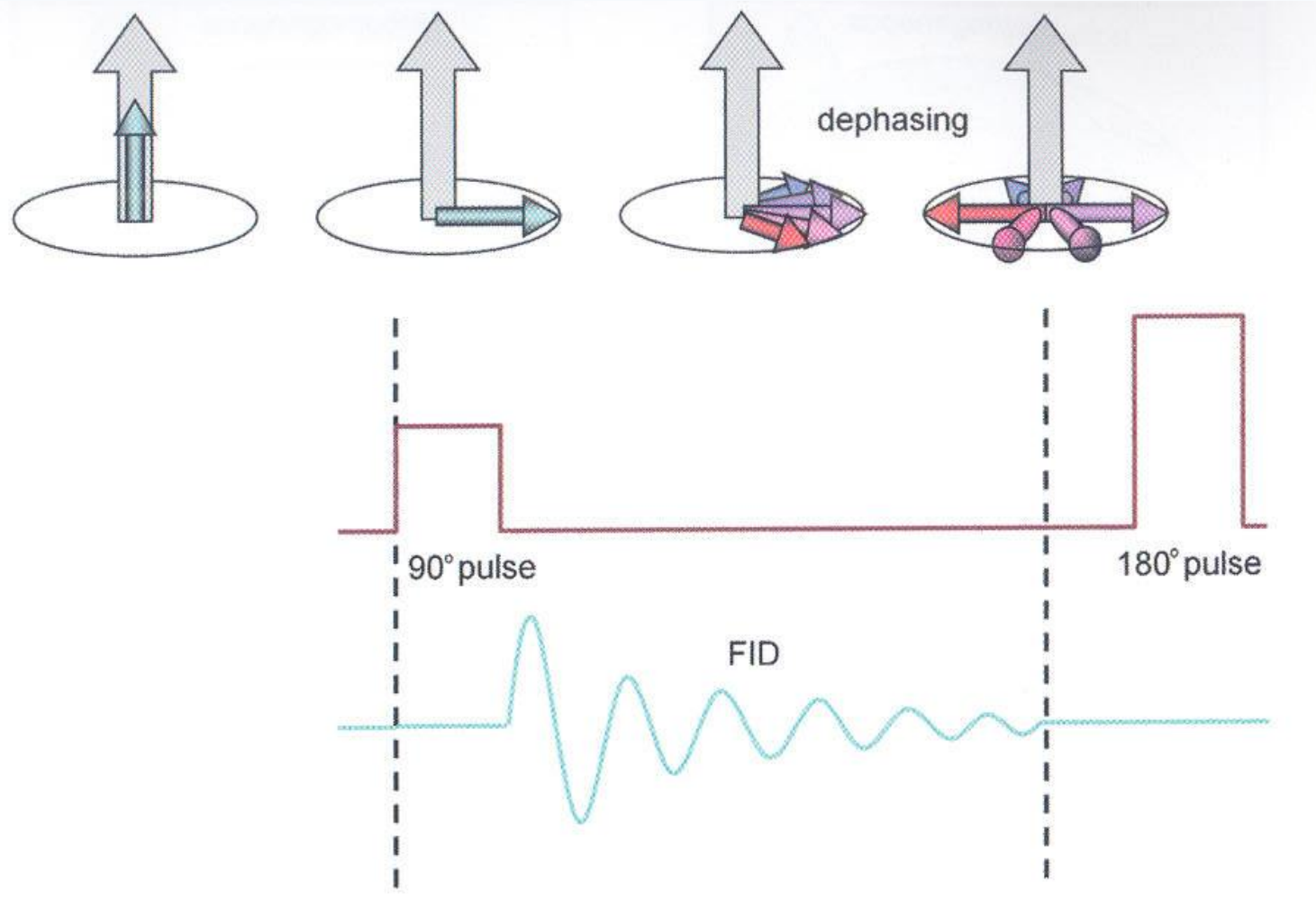
T2* relaxation

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
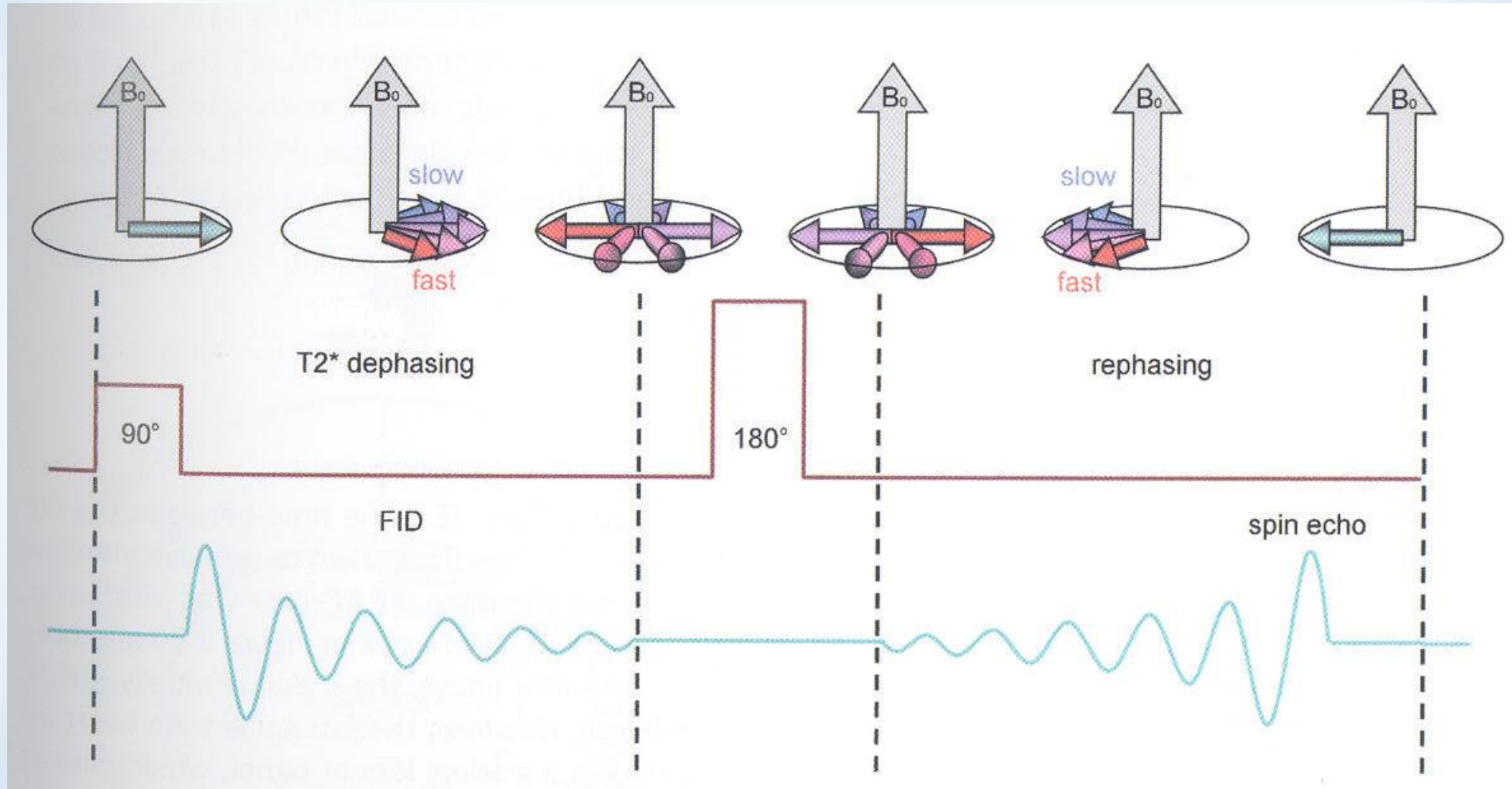
Rephase of spins after 180º RF- pulse in "rotating frame"

PRINCIPLE OF SPIN ECHO FORMATION

90º RF- pulse at t = - 0
phase-coherent spins

180º RF- pulse at t = τ

(a) M_{xy} t = 0

(b) F S t = τ

(c) M_{xy} t = 2τ
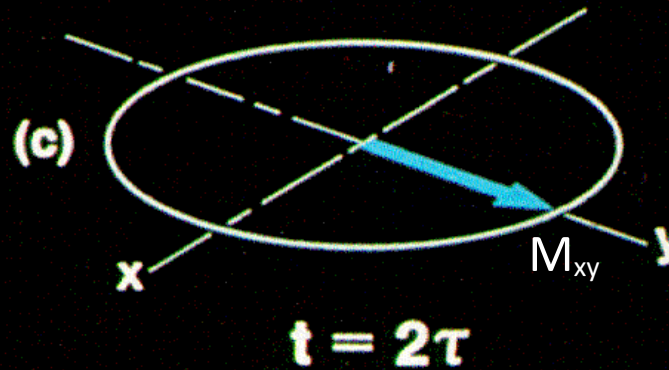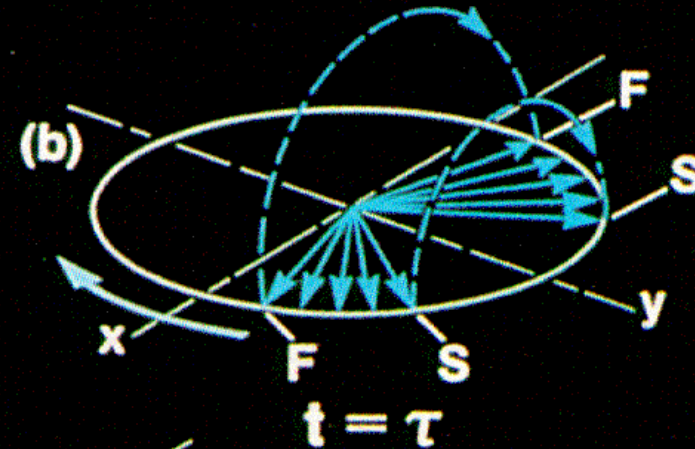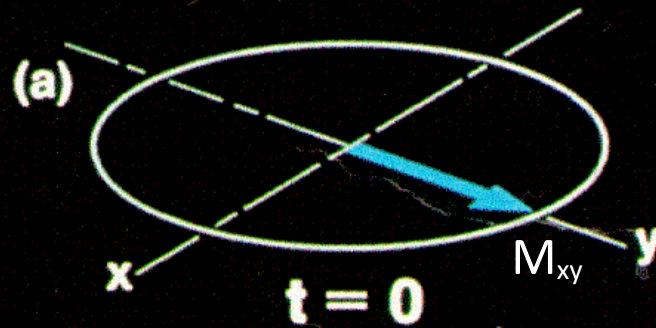
F = fast spins
S = slow spins

phase-coherent spins at t = 2τ

Effect of 180-pulse
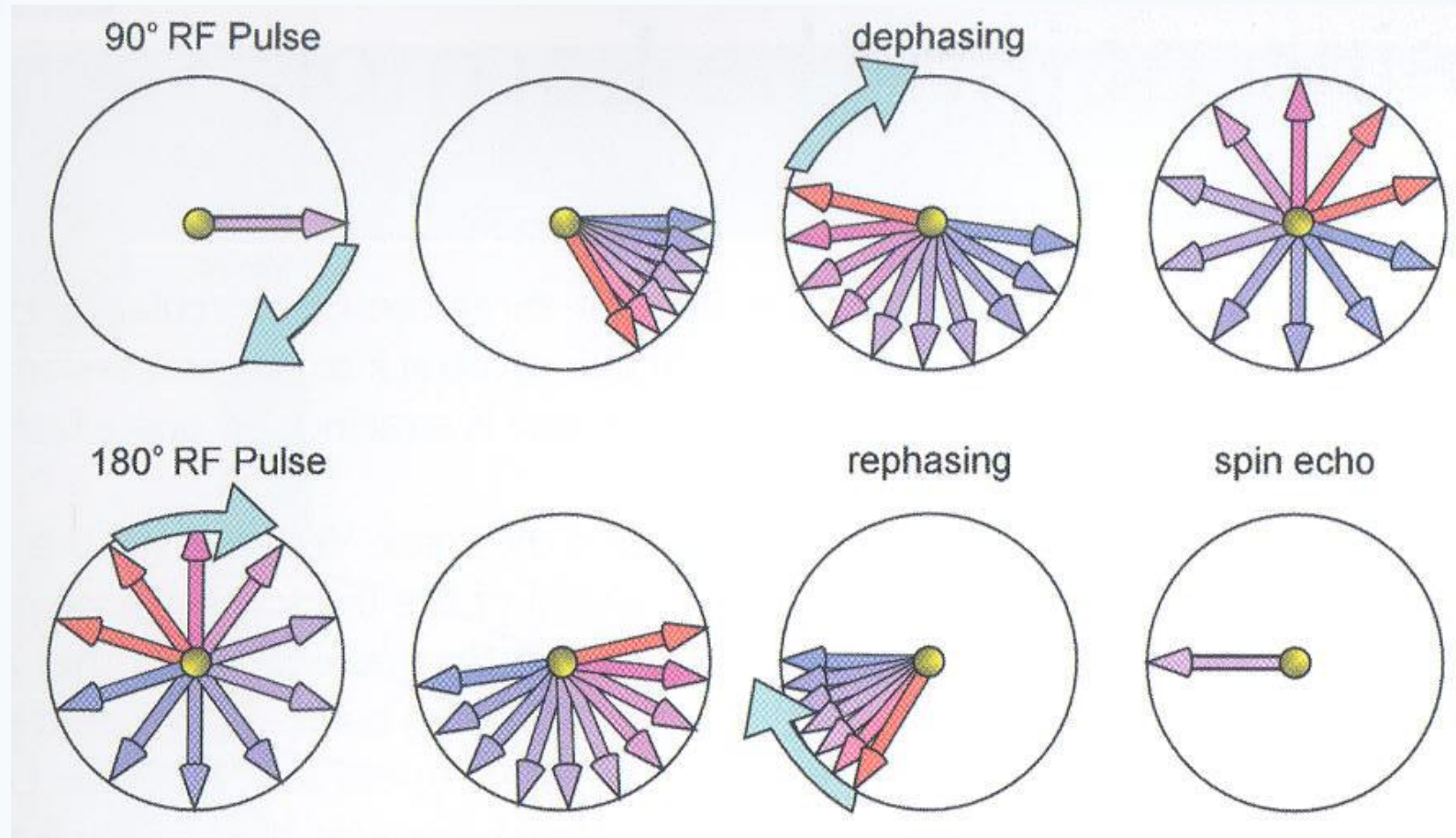
Relaxation

What happens to Magnetization after excitation?

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)

fat regains longitudinal magnetization rapidly

water regains longitudinal magnetization slowly
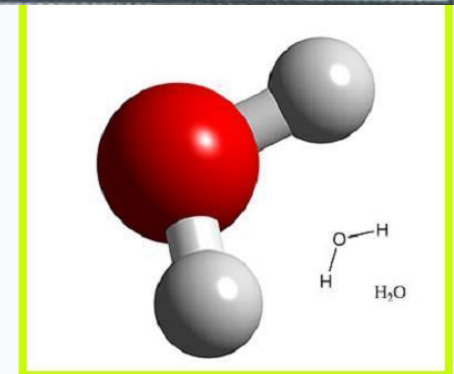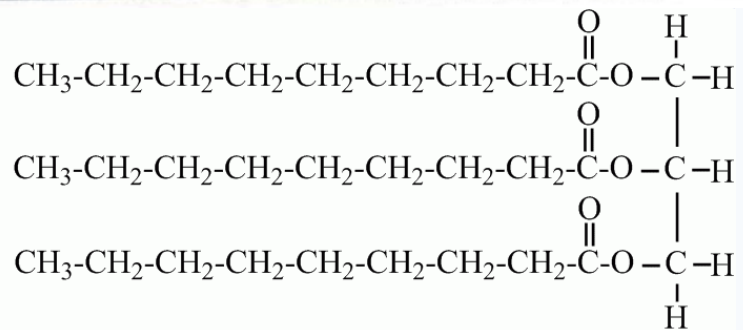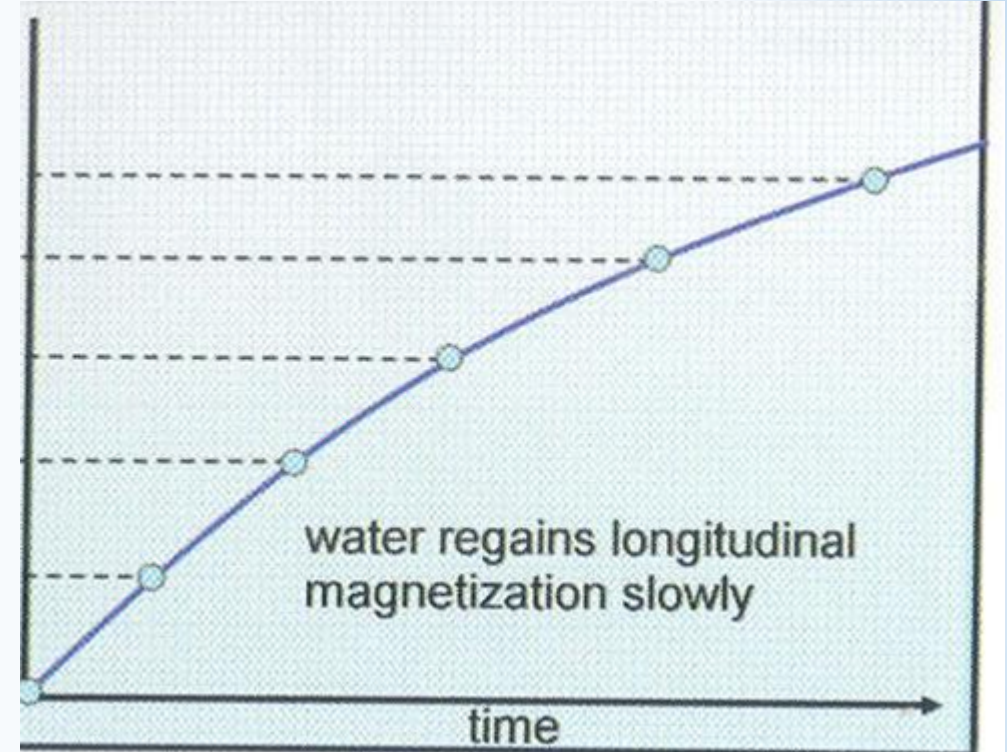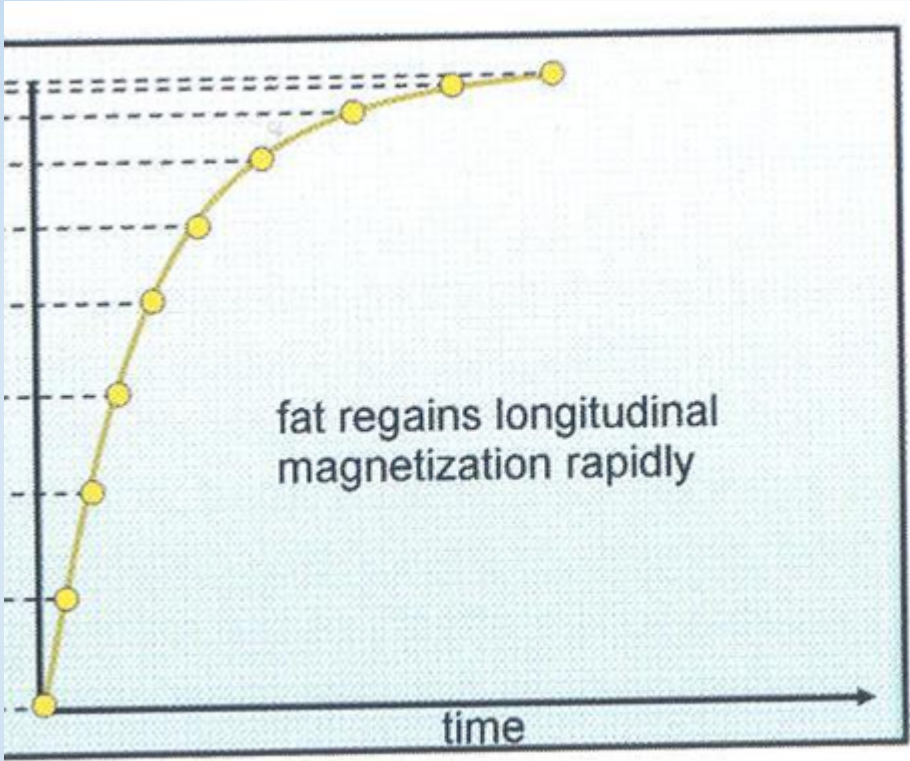
T1 relaxation time

Tissue T_1-relaxation time 400 ms
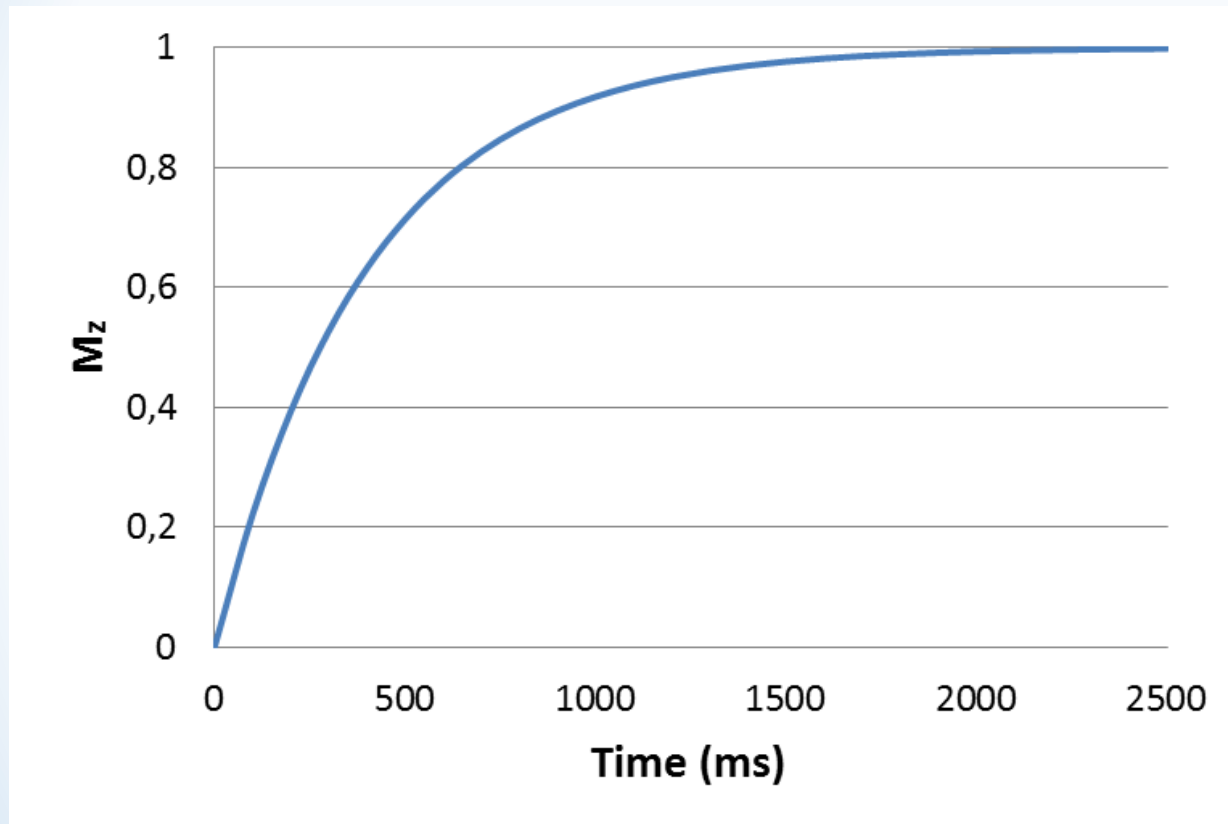
TR (ms)	M_z
200	39,3%
400	63,2%
800	86,5%
1200	95,0%
2000	99,3%

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

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)
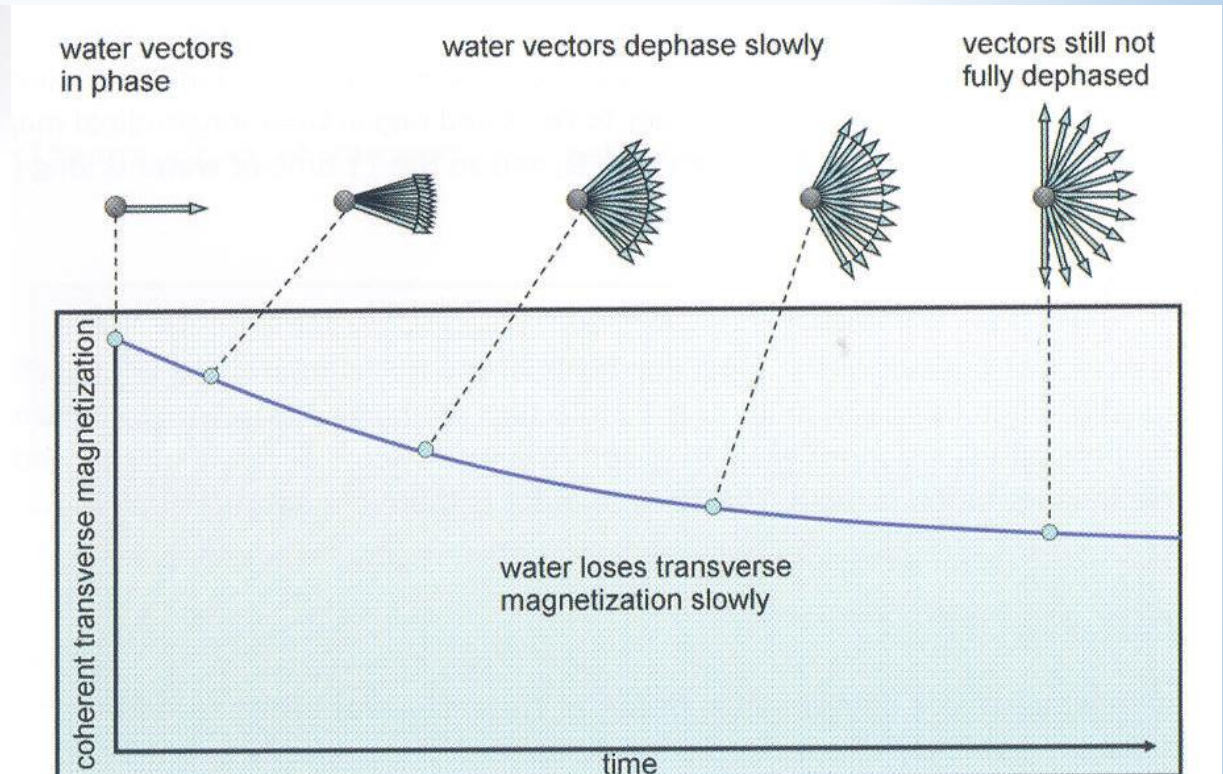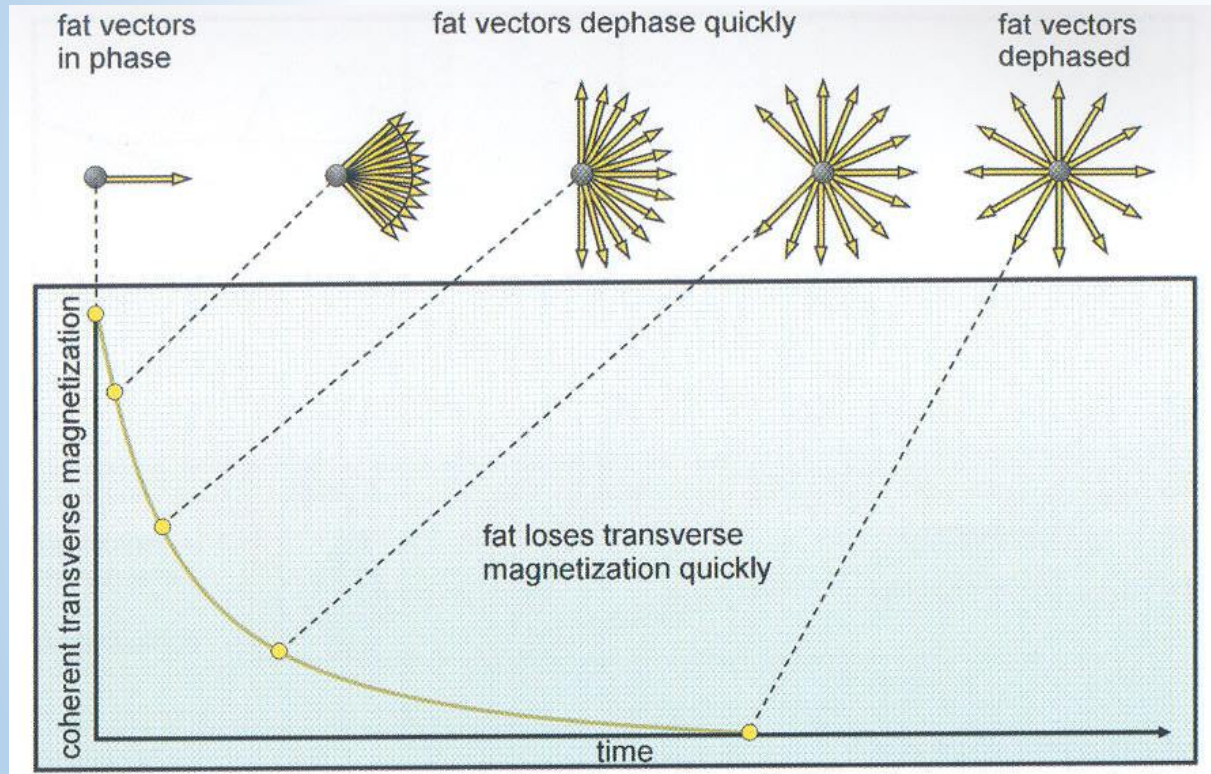
T2 relaxation time

Tissue T_2-relaxation time 40 ms
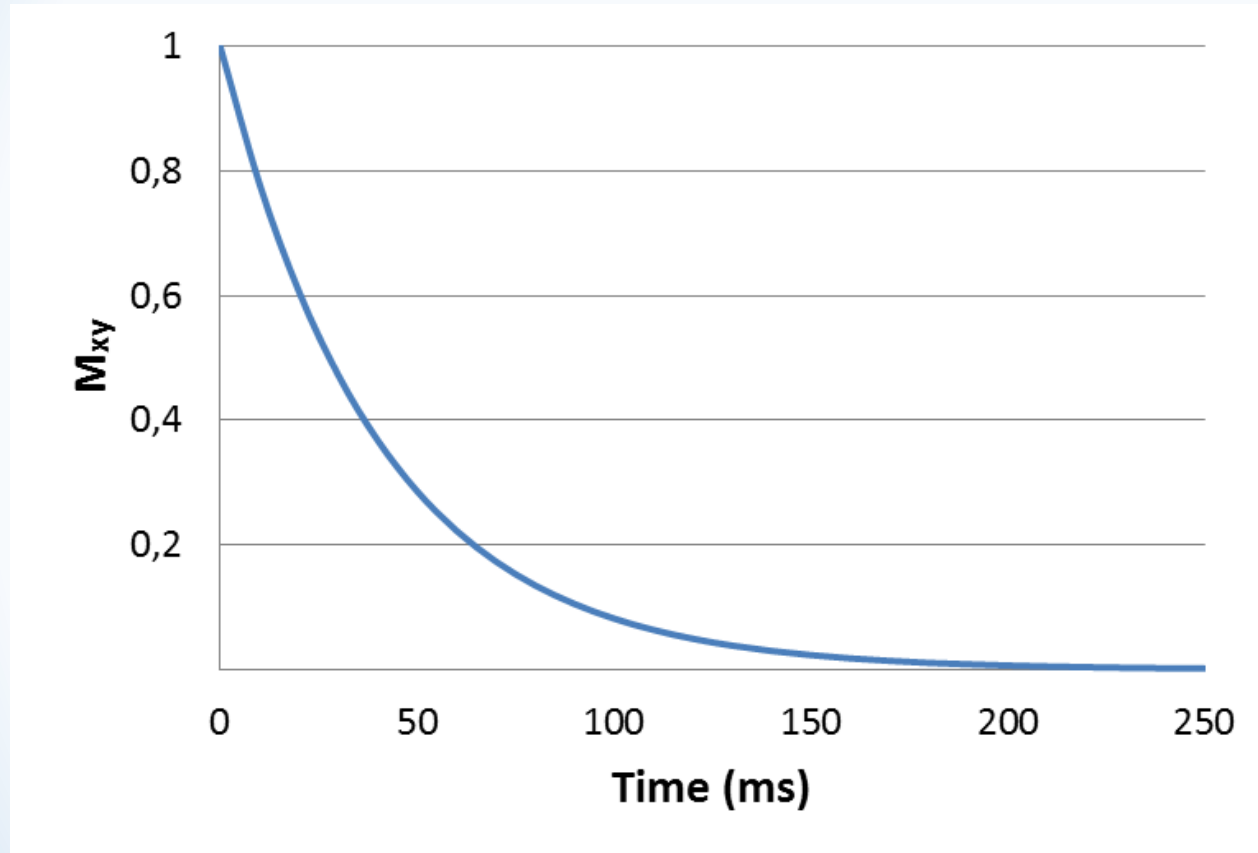
TE(ms)	M_{xy}
20	60.7%
40	36,8%
80	13,5%
120	5,0%
200	0,7%

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

Contrast
Pulse sequence (simplest case)
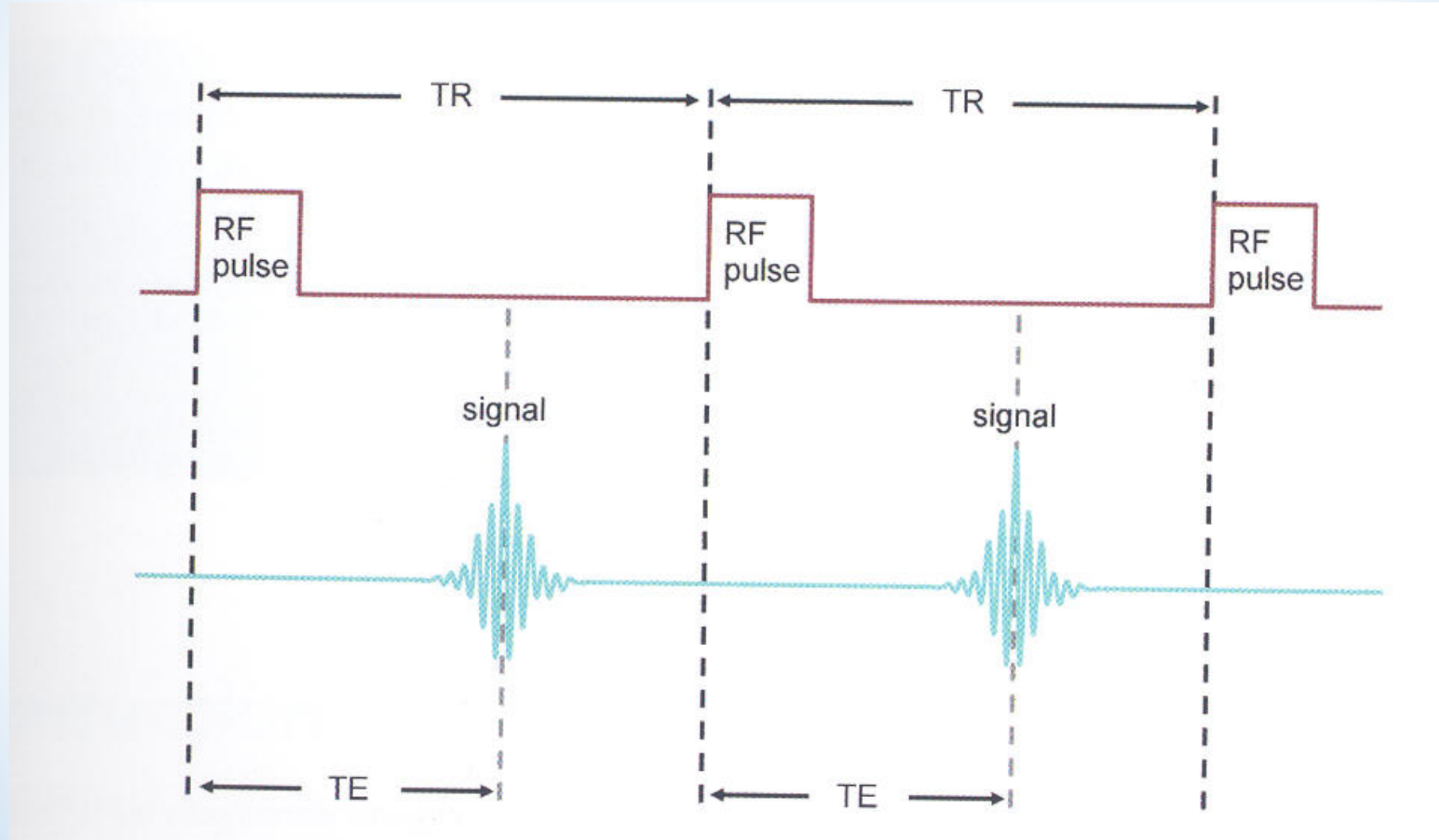
Contrast
Pulse sequence – Spin Echo sequence (simple case)
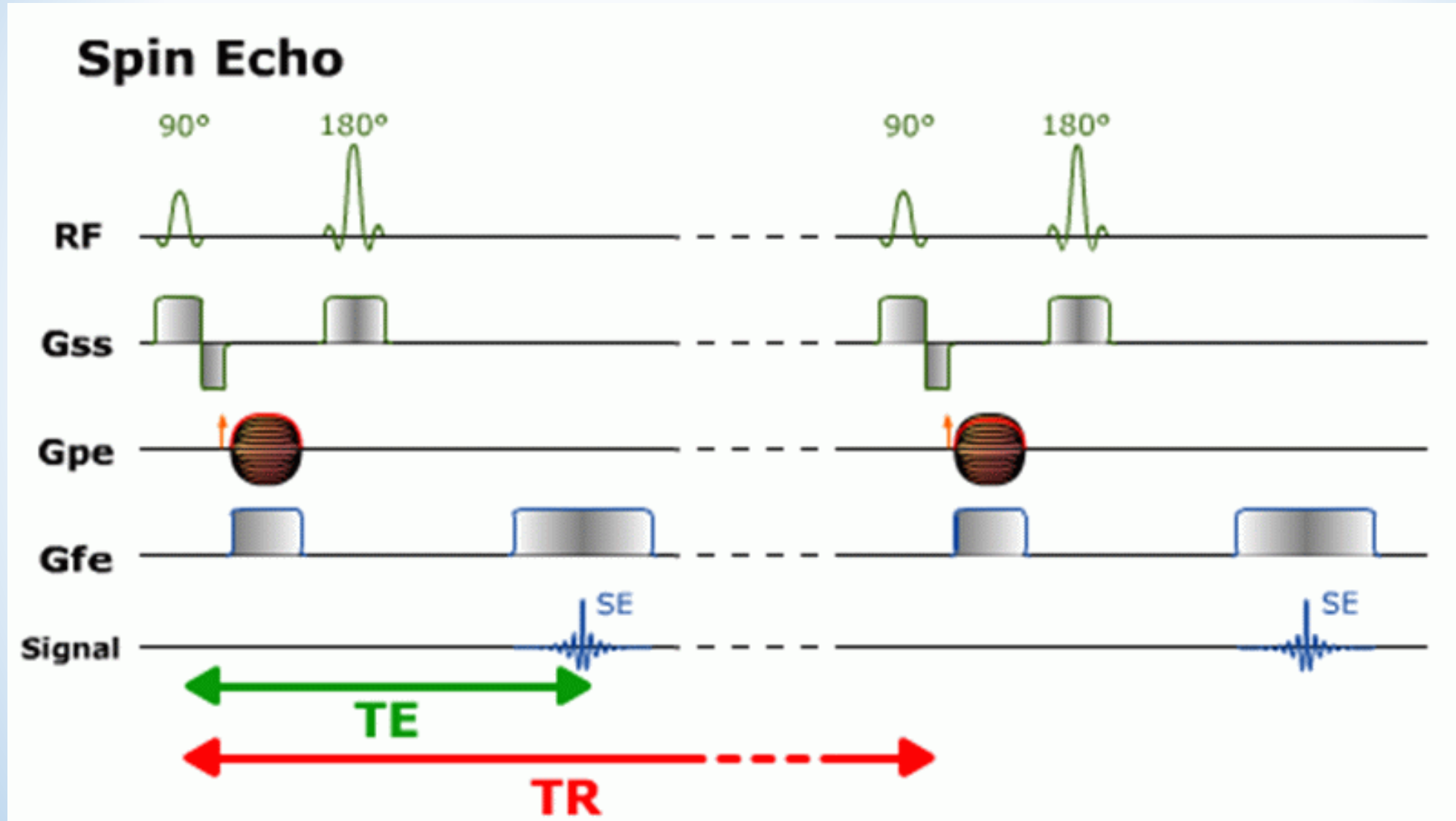
Image contrast and contrast agents

T1-contrast
(fat vs water)
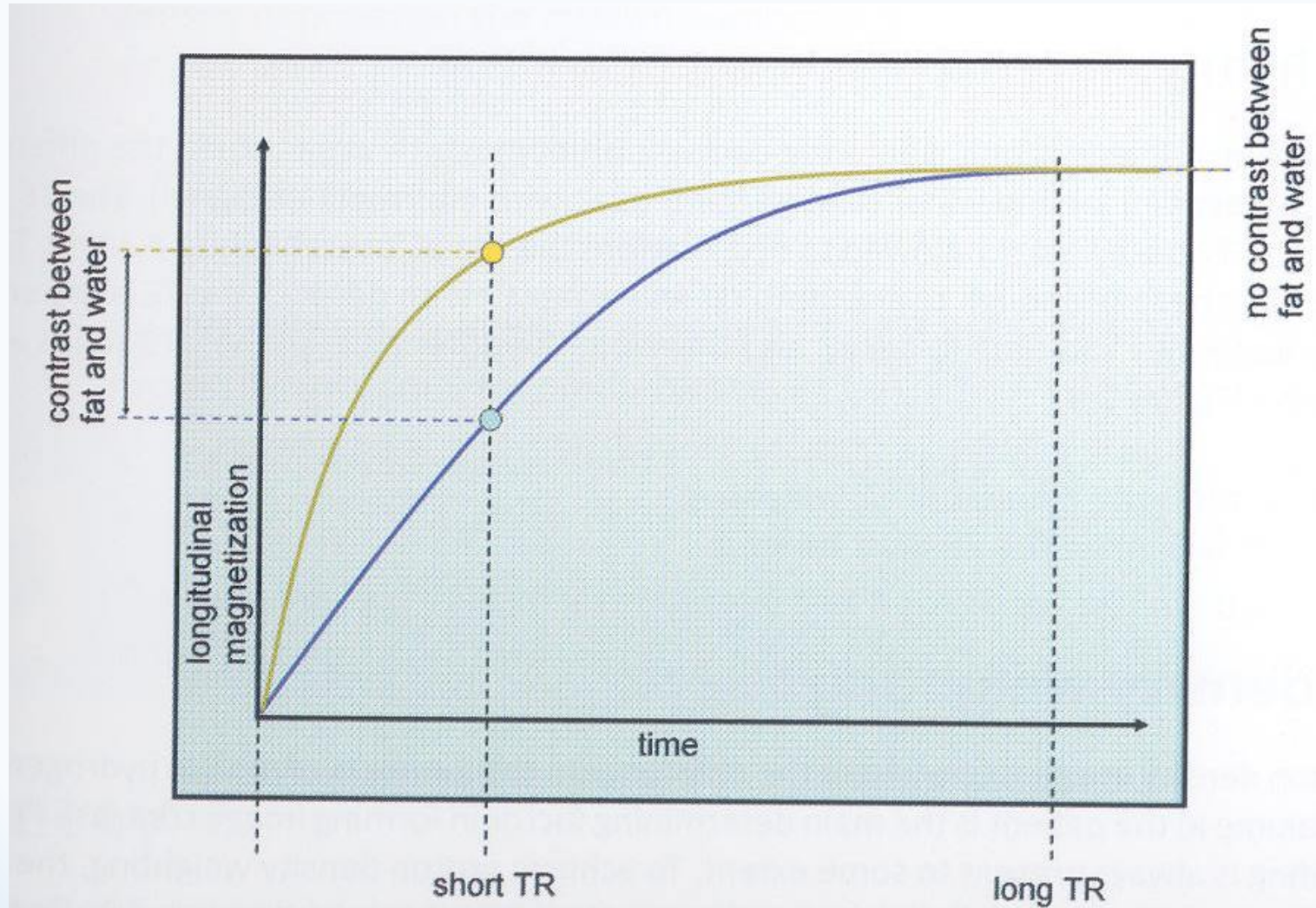
T2-contrast
(fat vs water)
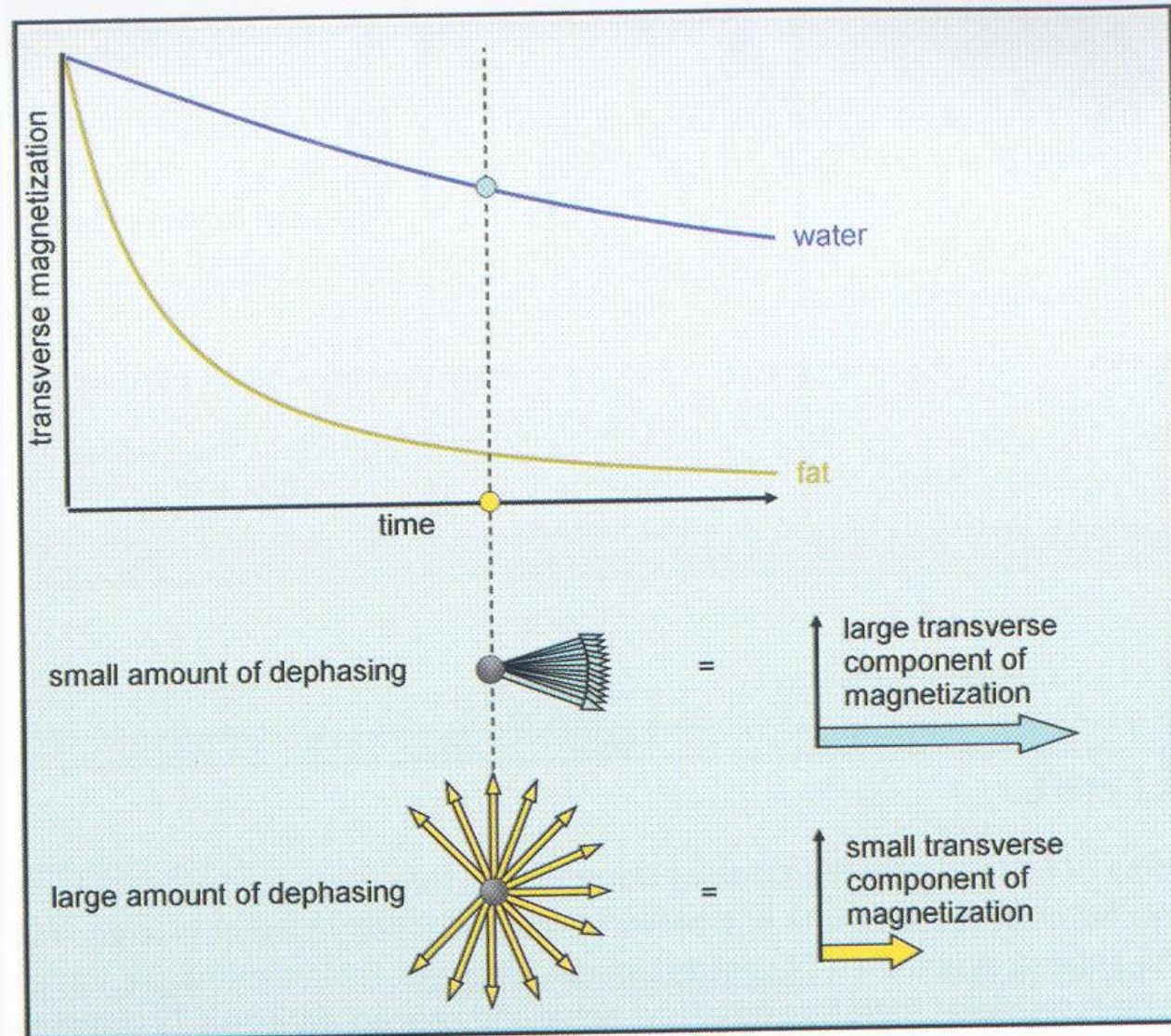
T2-contrast
(fat vs water)
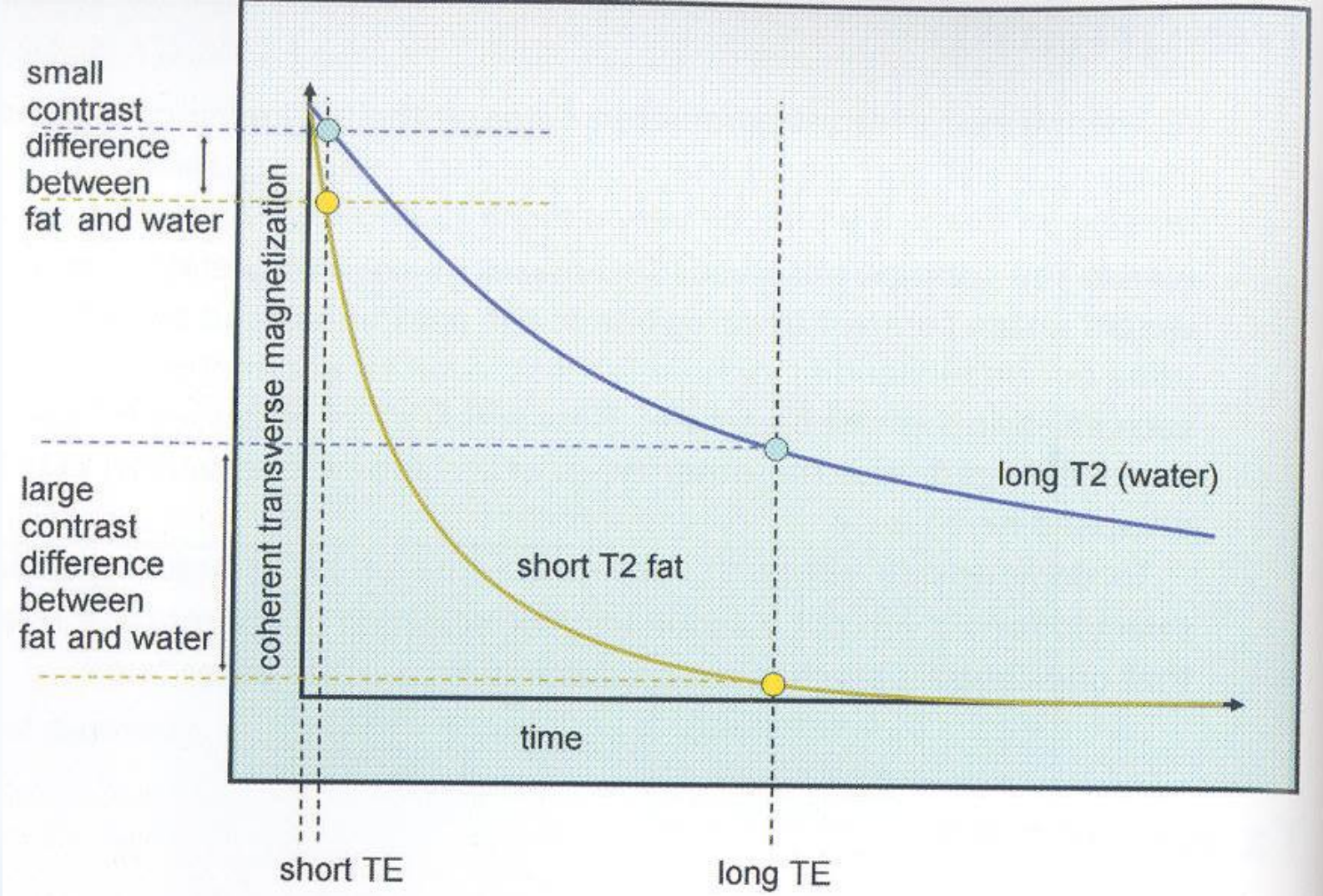
Relaxation times in tissue

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

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
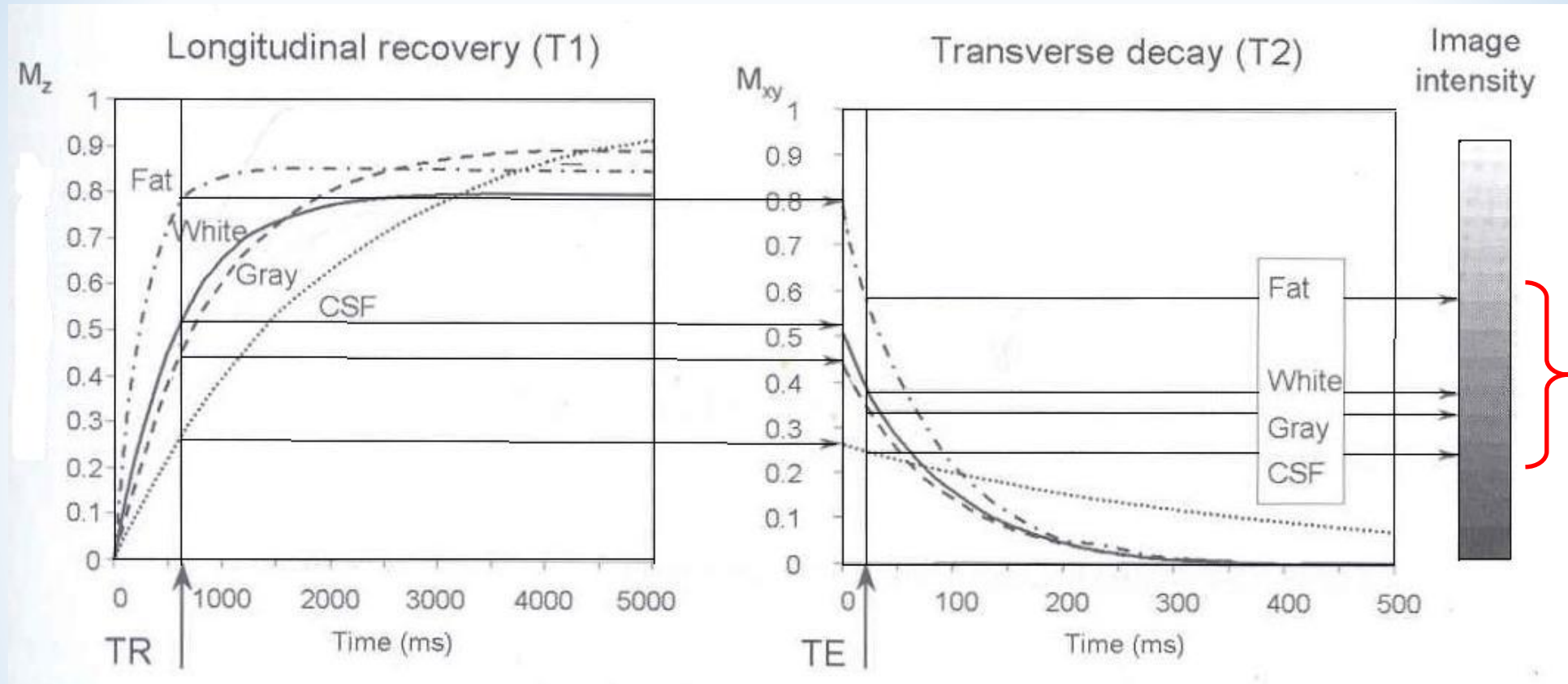
Short TR (300 – 700 ms)

Short TE (5 – 25 ms)

Fat is "bright" (high signal), CSF "dark" (low signal)
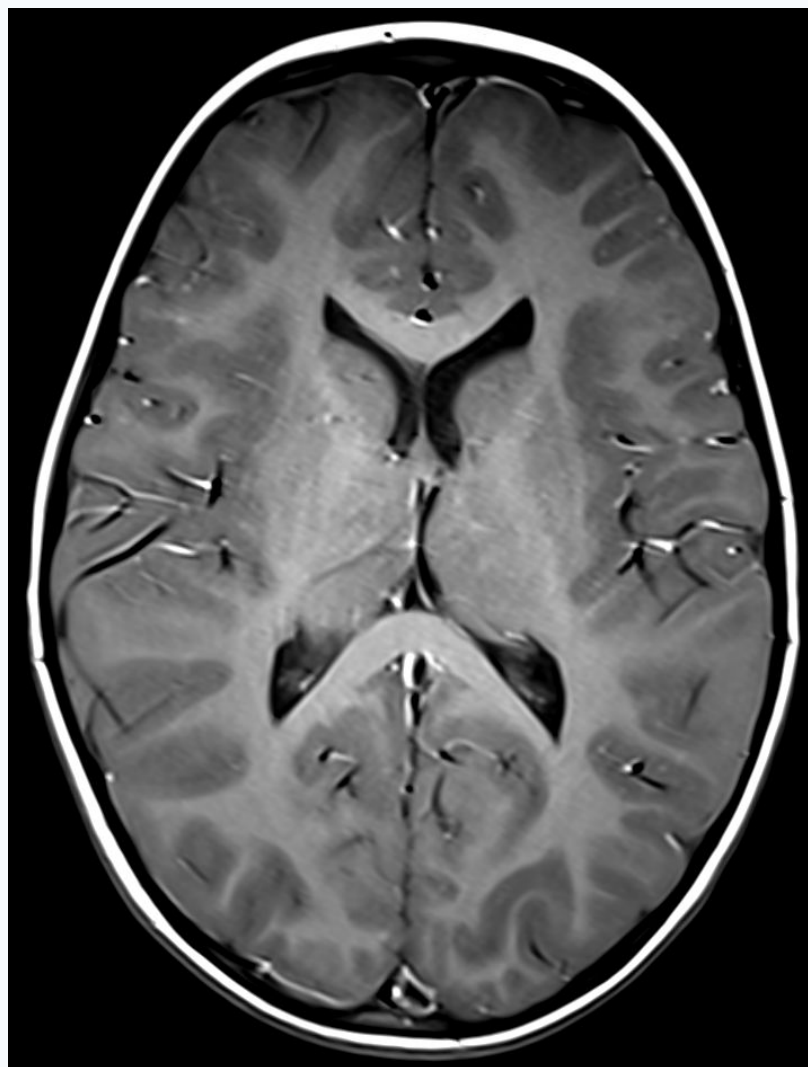
T1-weighted image
TR = 600 ms, TE = 6,4 ms

Proton density (PD) weighted scan
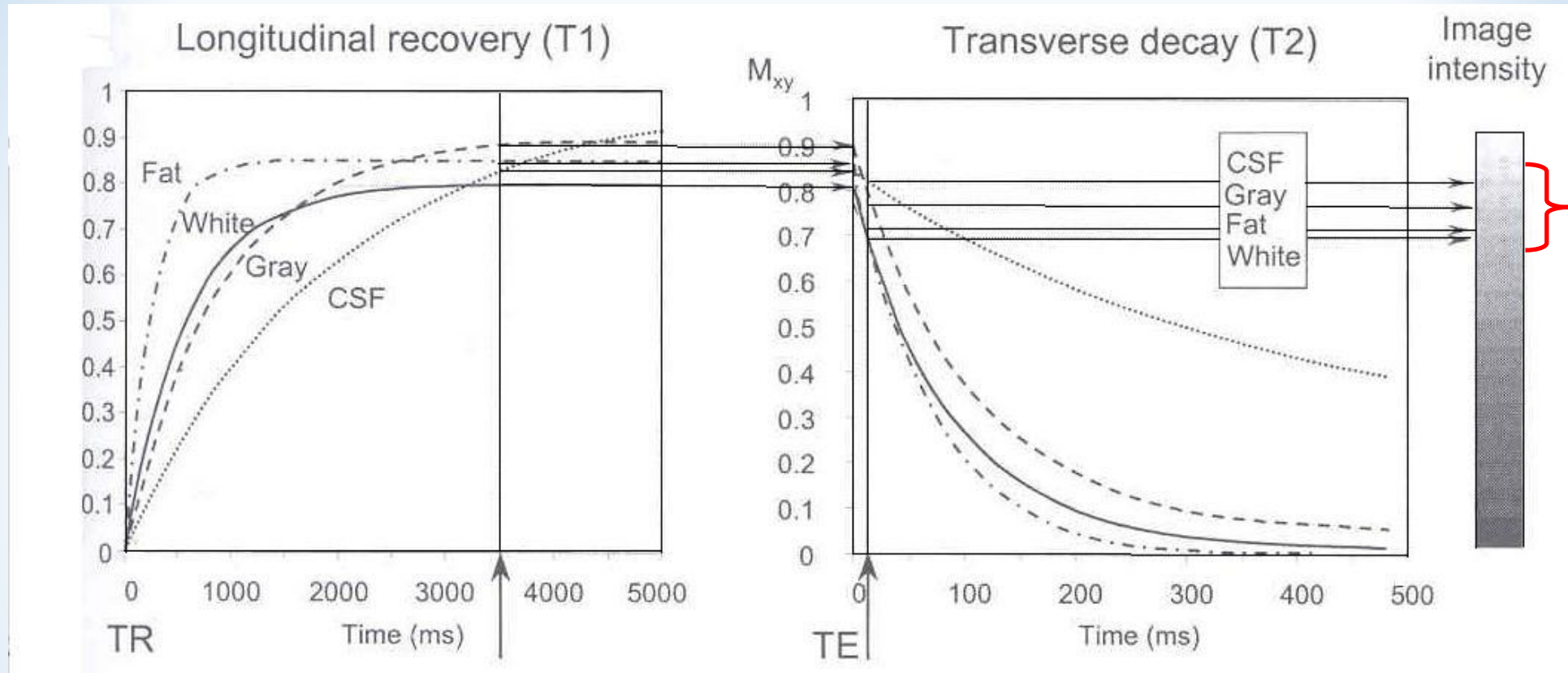
Long TR (2 – 4 s)

Short TE (5 – 30 ms)

Highest signal, but often weak contrast
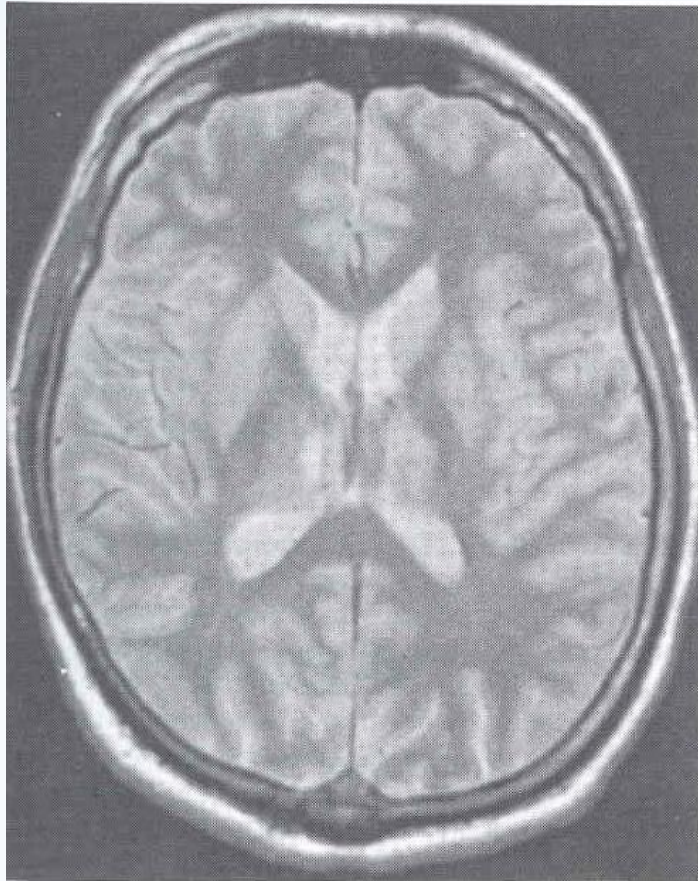
PD-weighted

TR = 2400 ms, TE = 30 ms
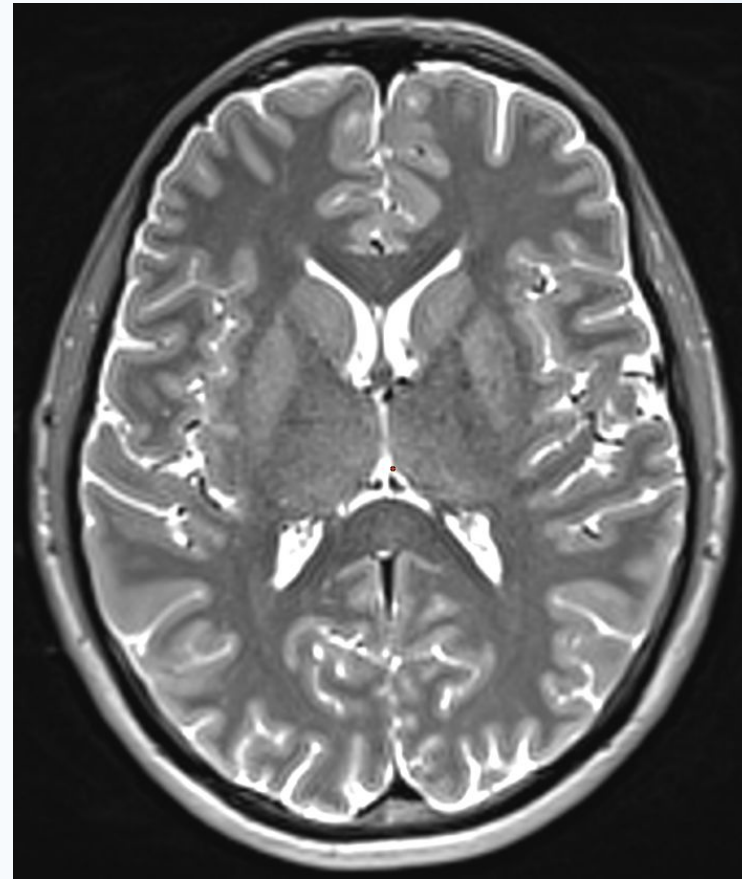
PD-weighted TSE

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

T2-weighted scan
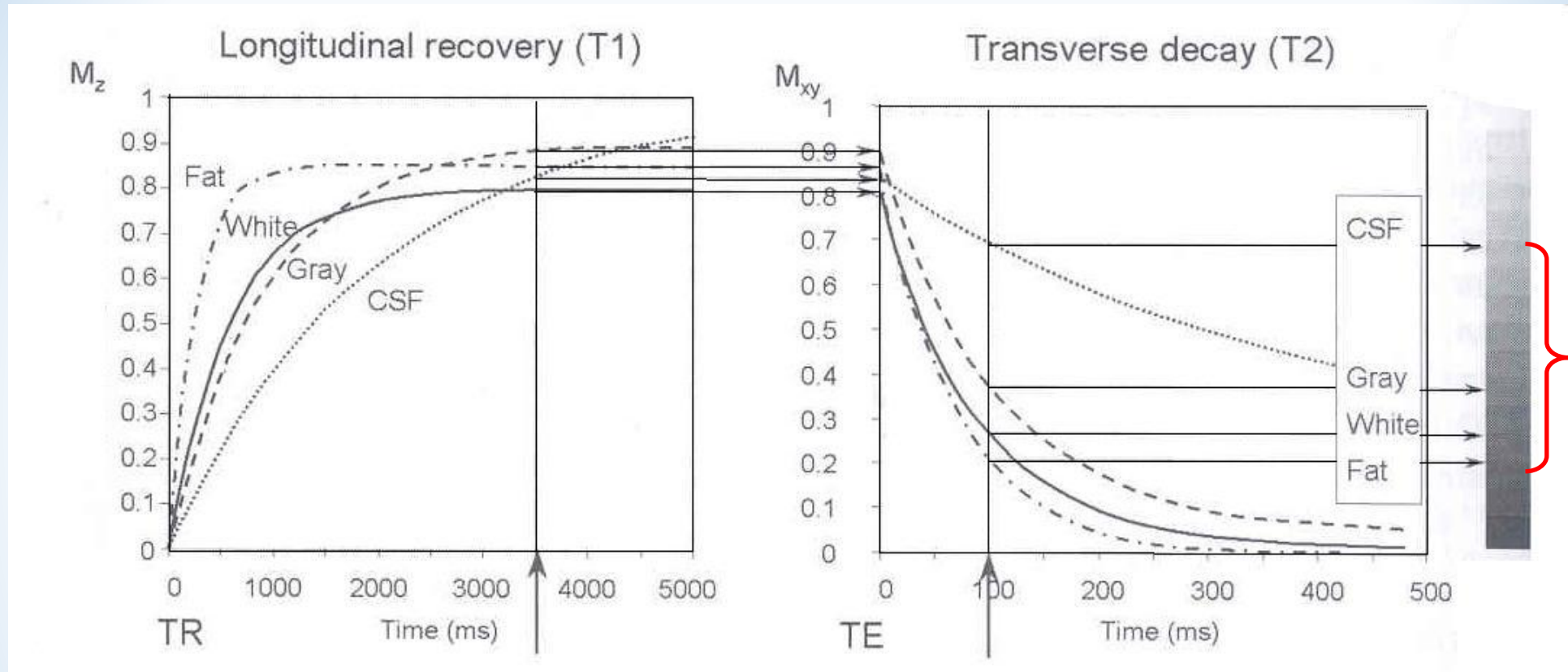
Long TR (2 – 4 s)

Long TE (60 – 120 ms)

CSF ”bright” (strong signal), fat ”dark” (weak signal)
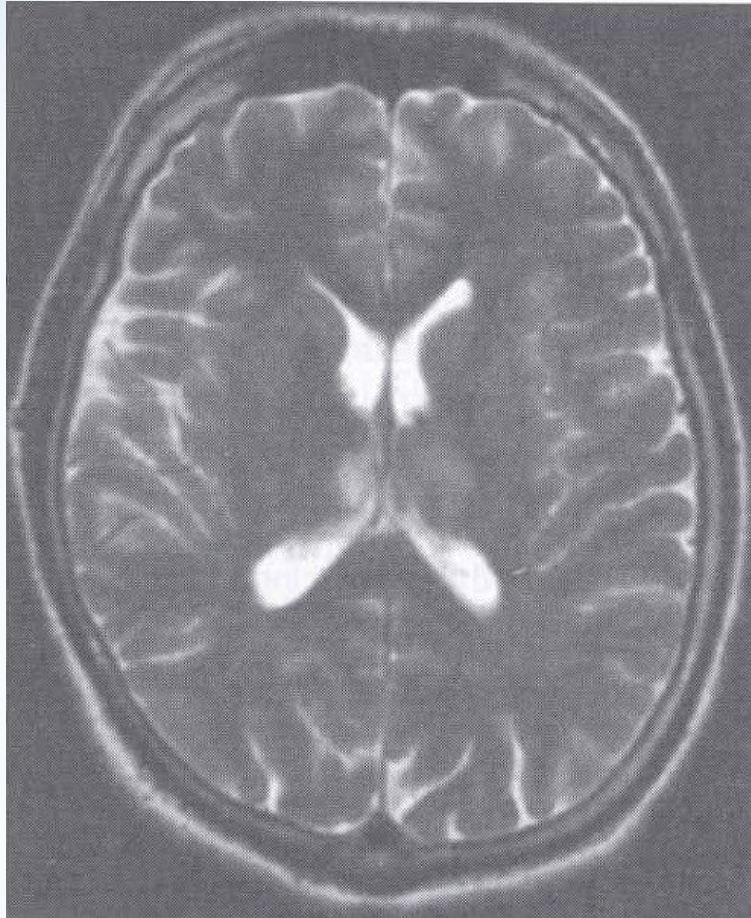
T2-weighted SE

TR = 2400 ms, TE = 90 ms
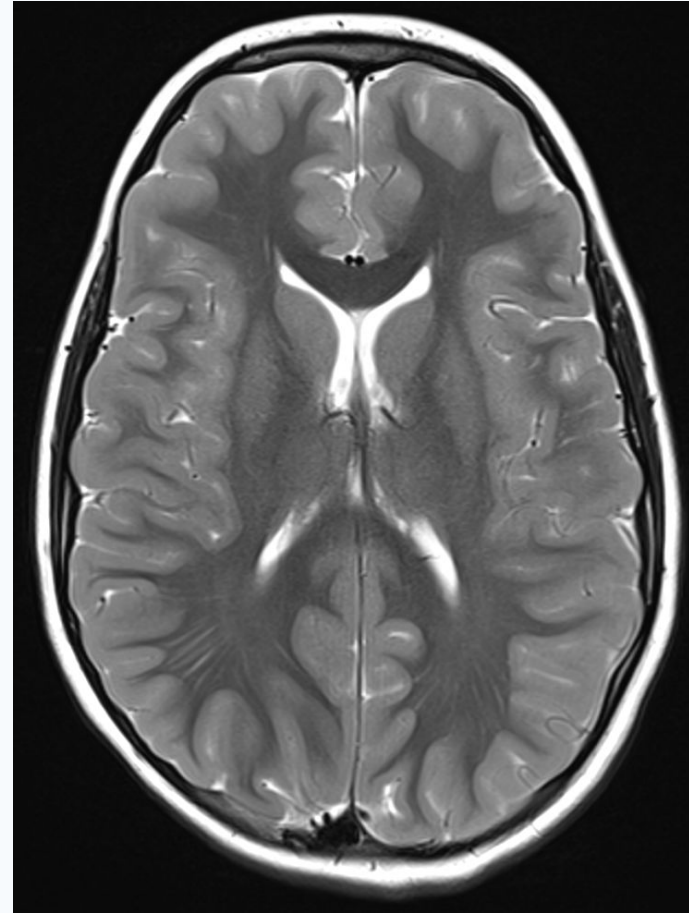
T2-weighted TSE

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

T2 contrast in head MR images

TR = 2 500
TE =   15

TR = 2 500
TE =   60

TR = 2 500
TE =   90

TR = 2 500
TE =   120
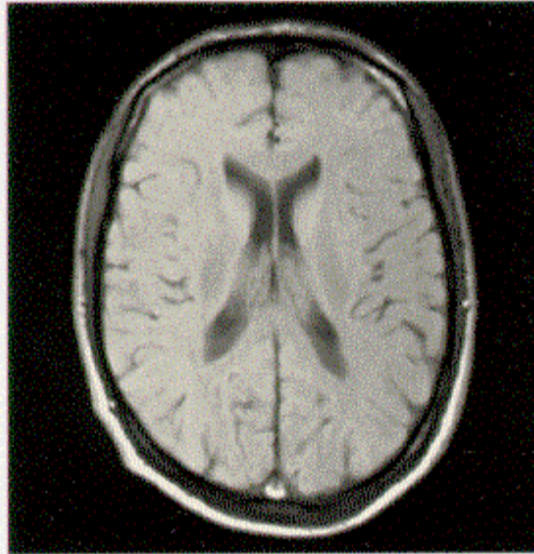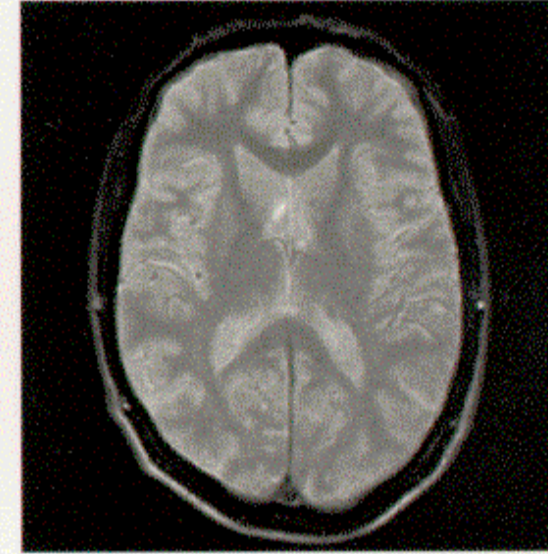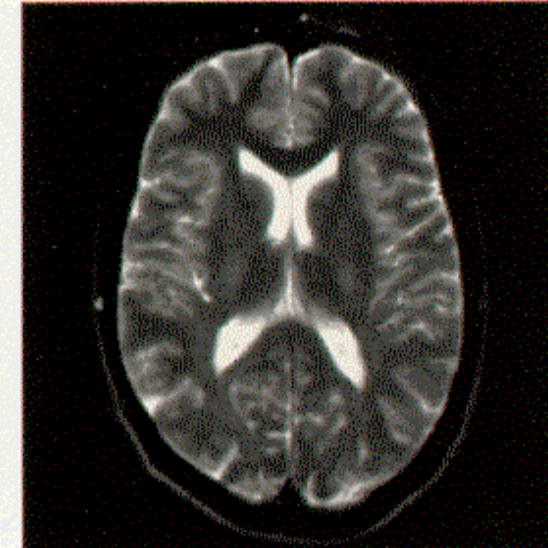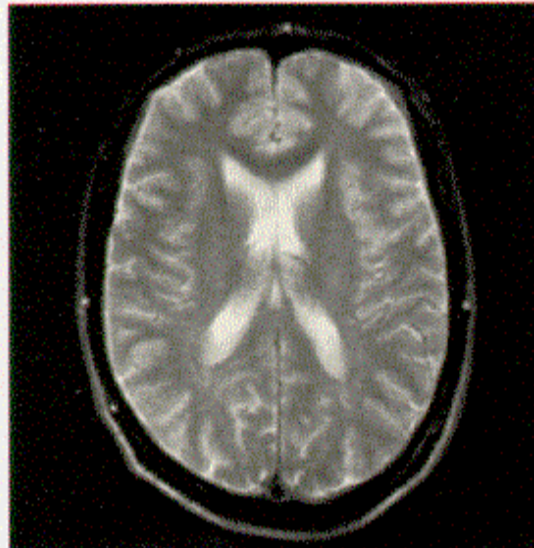
1: Proton density- weighted (PDW) image

2: Moderately T2- weighted (T2W) image

3: Routine T2- weighted (T2W) image

4: Heavily T2- weighted (T2W) image

Effect of TR and TE

TE = 20
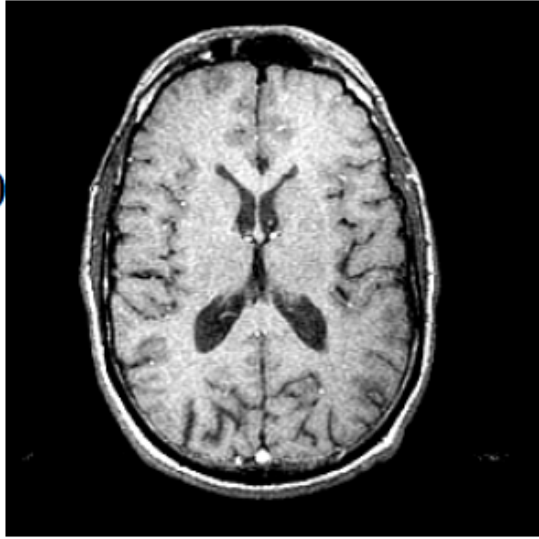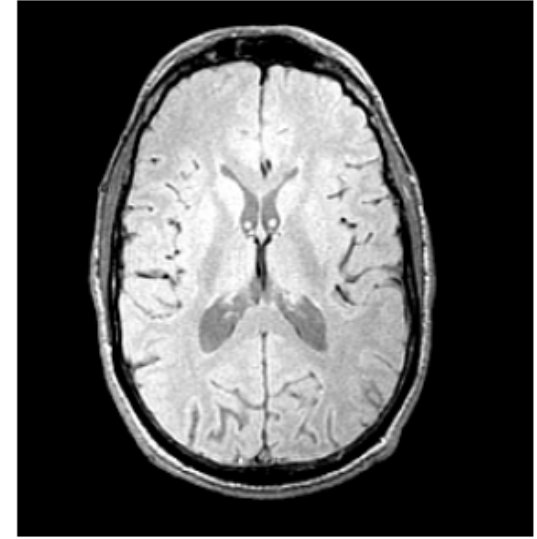
TR = 500   TR = 1000   TR = 2000

TE = 80
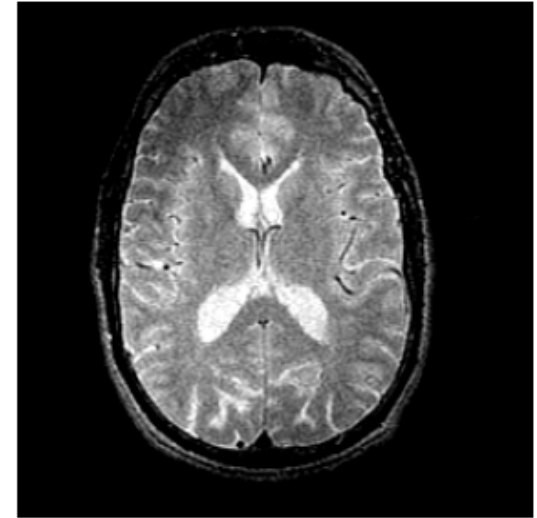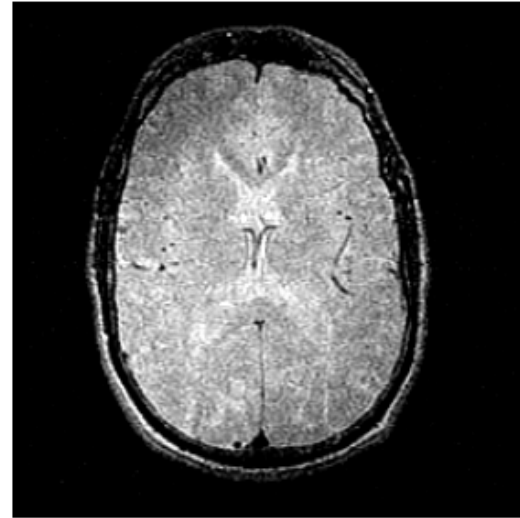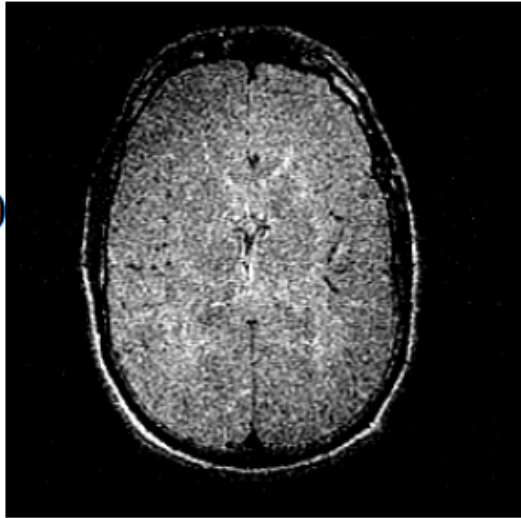
Contrast agents

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
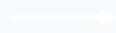
MRI contrast in brain: an atypical meningioma

PD- weighted,
PDW
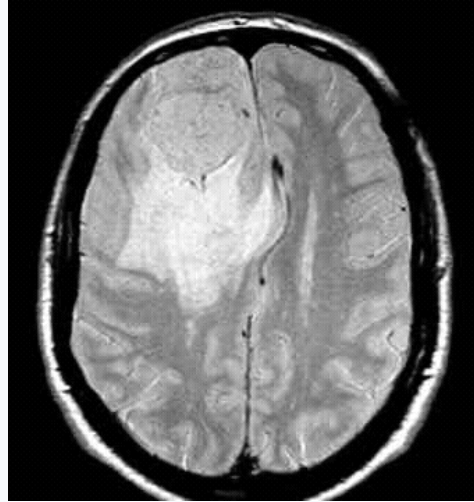
T2-weighted,
T2W
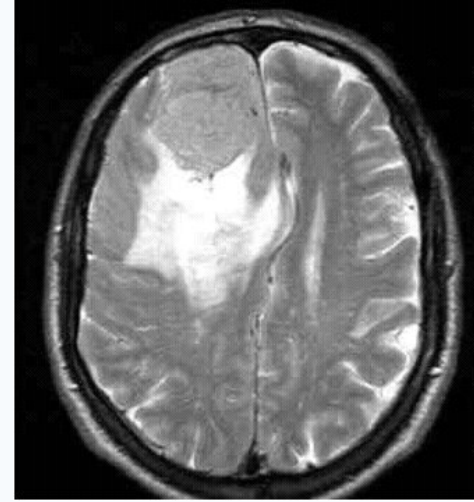
T1-weighted,
T1W,
no contrast
agent
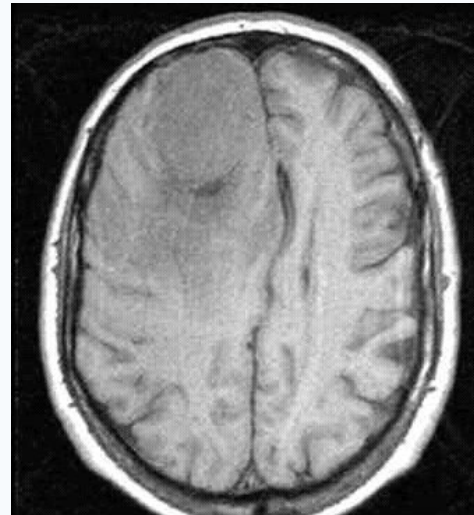
T1-weighted,
T1W,
With contrast
agent

Dynamical CE Breast MRI
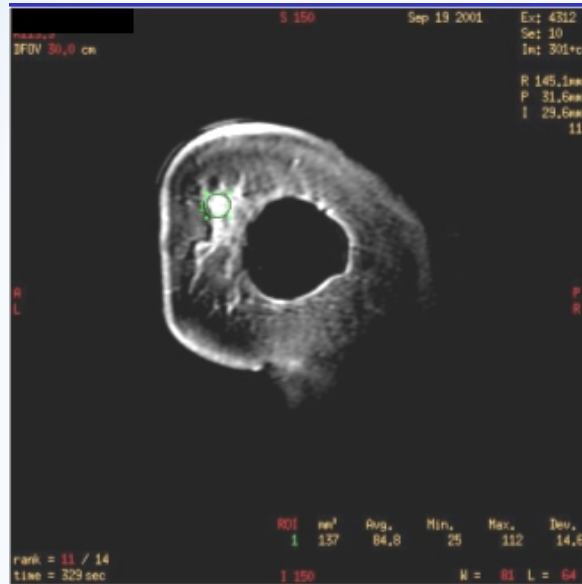
Gd-DTPA- enhanced magnitude image
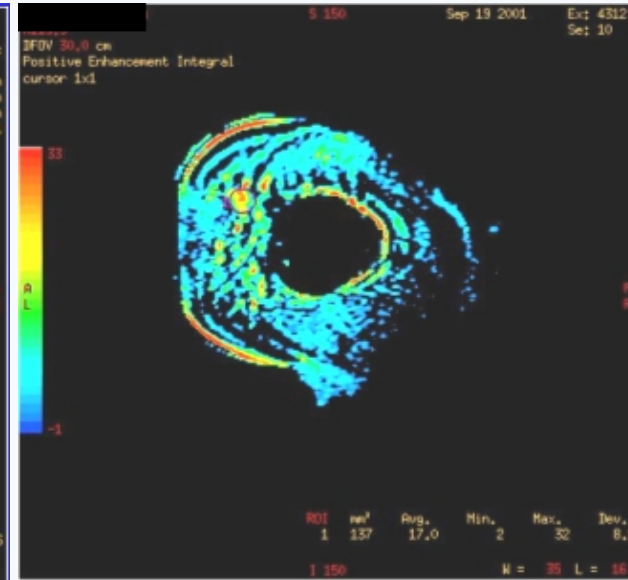
Positive Enhancement Integral
PEI map
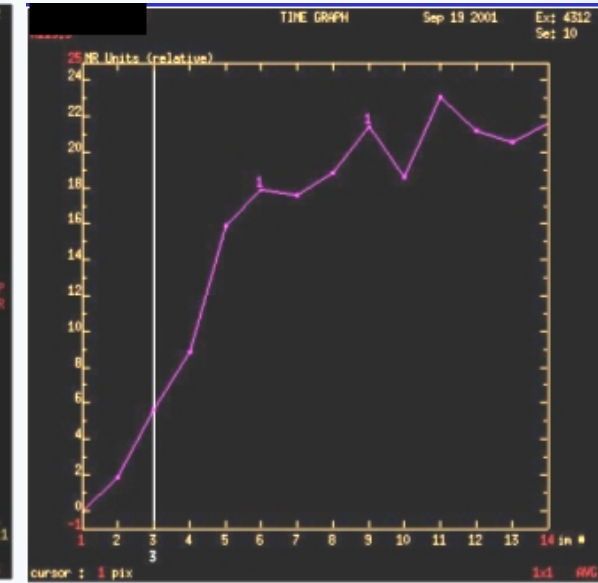
Intensity enhancement dynamical curve

From Signal to Image

GRADIENTS

Magnetic field gradient

Linearity of the gradient extremely important!
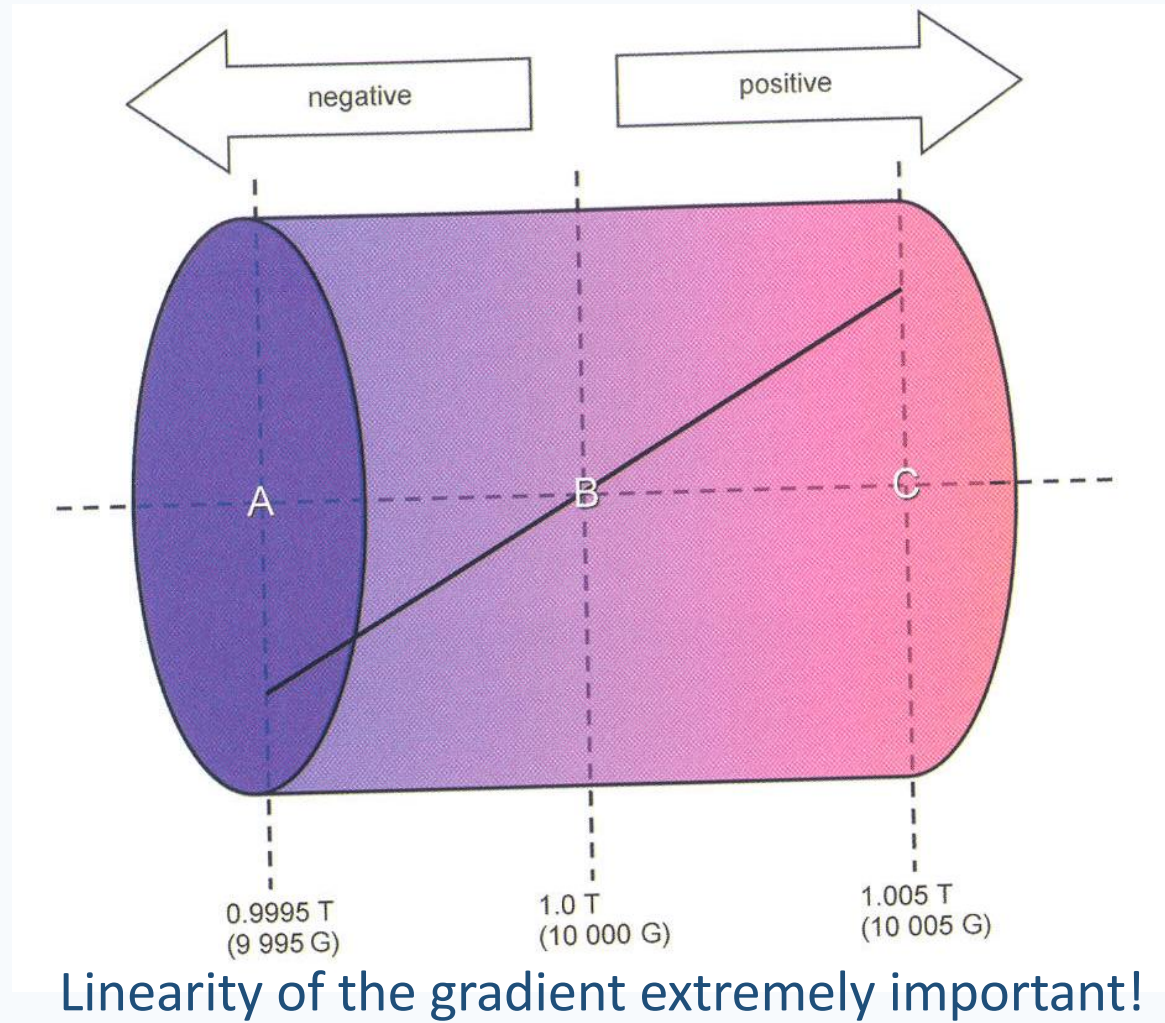
Gradient axis directions

Gradient axes in a typical superconducting system.

MRI Scanner Cutaway

Helium dewar

Superconducting coil

Gradient coils

Shim coils

B_0

RF coils
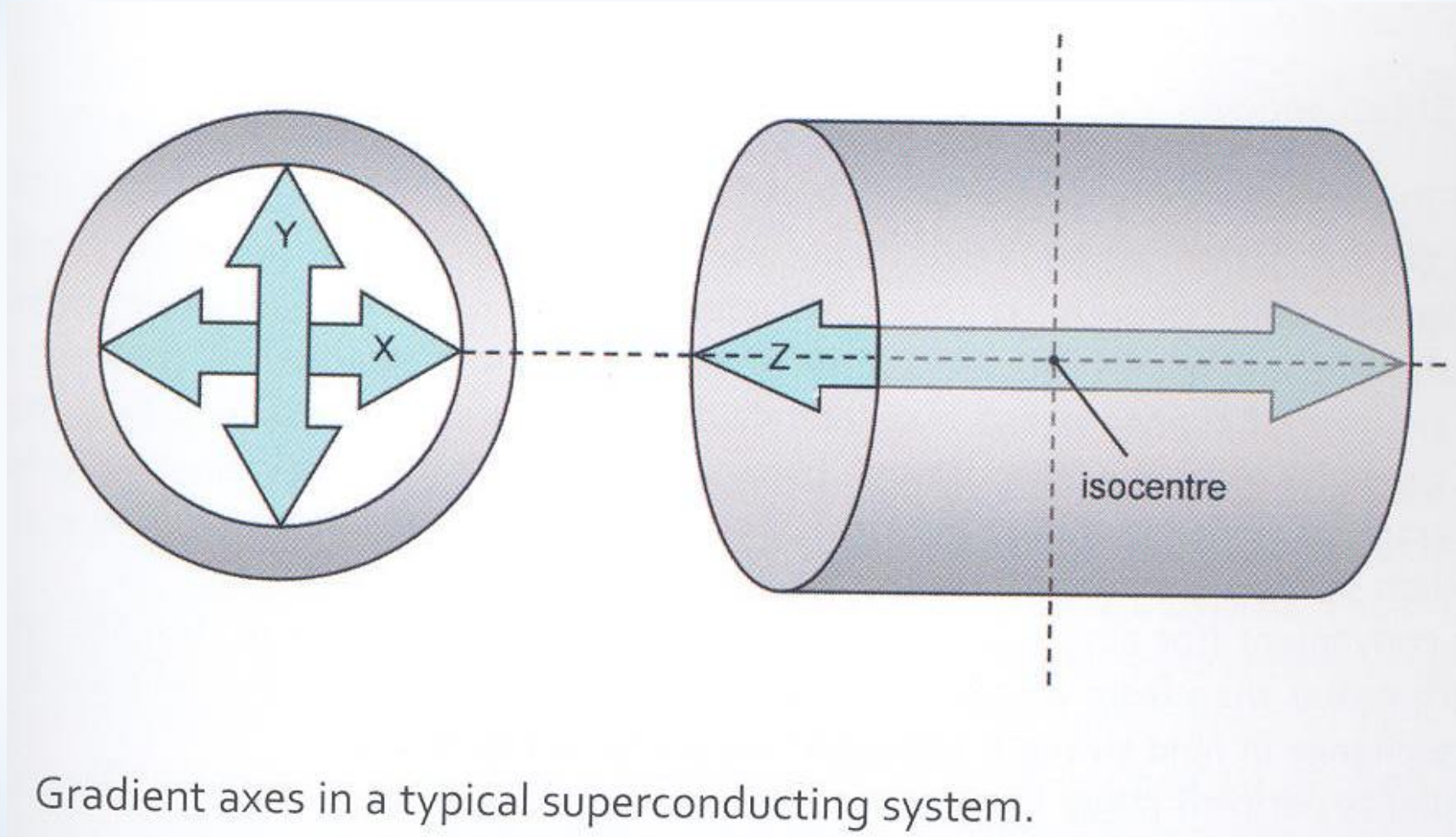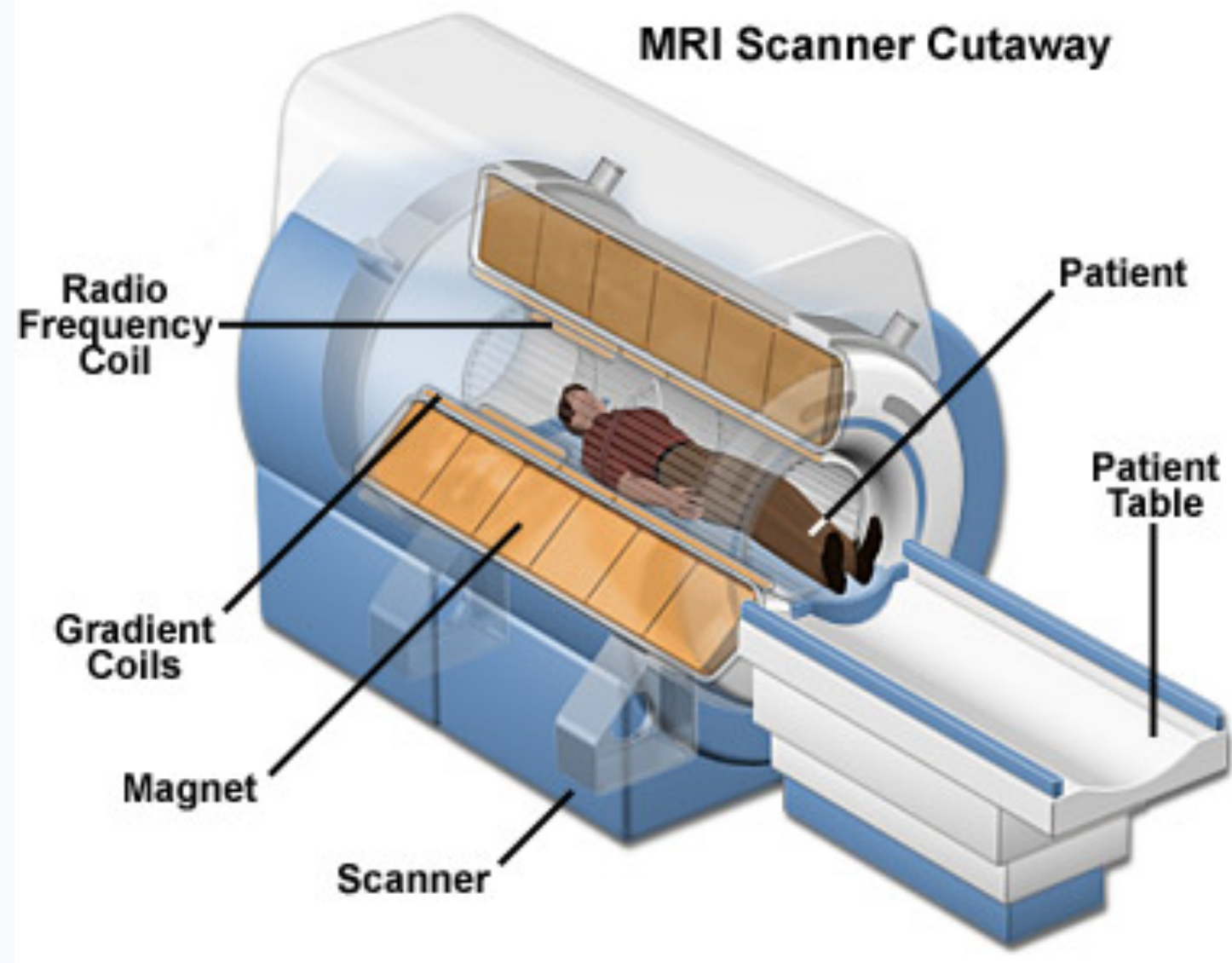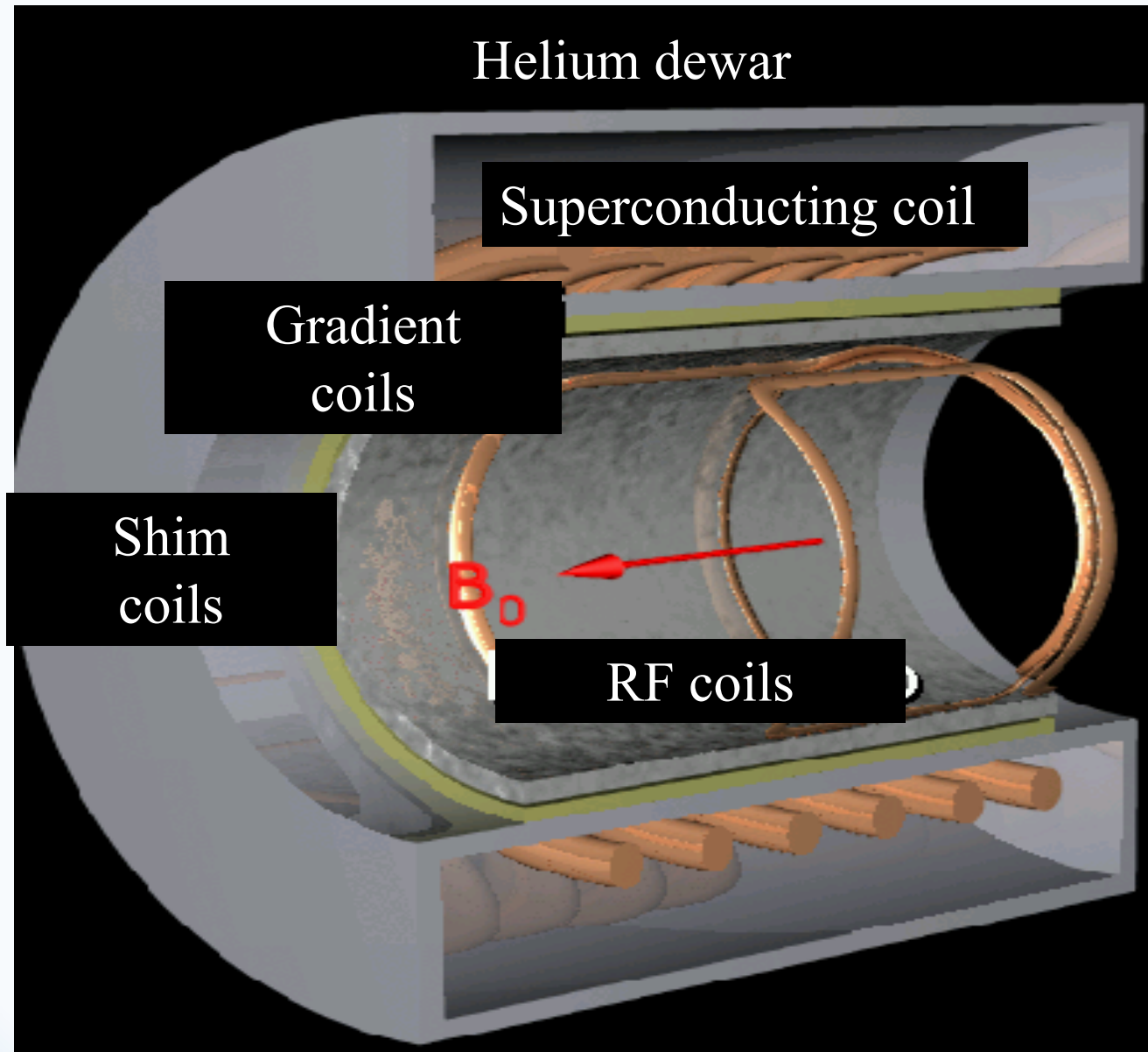
MRI Scanner Gradient Magnets

Y Coil

Z Coil

X Coil

Transceiver

Patient
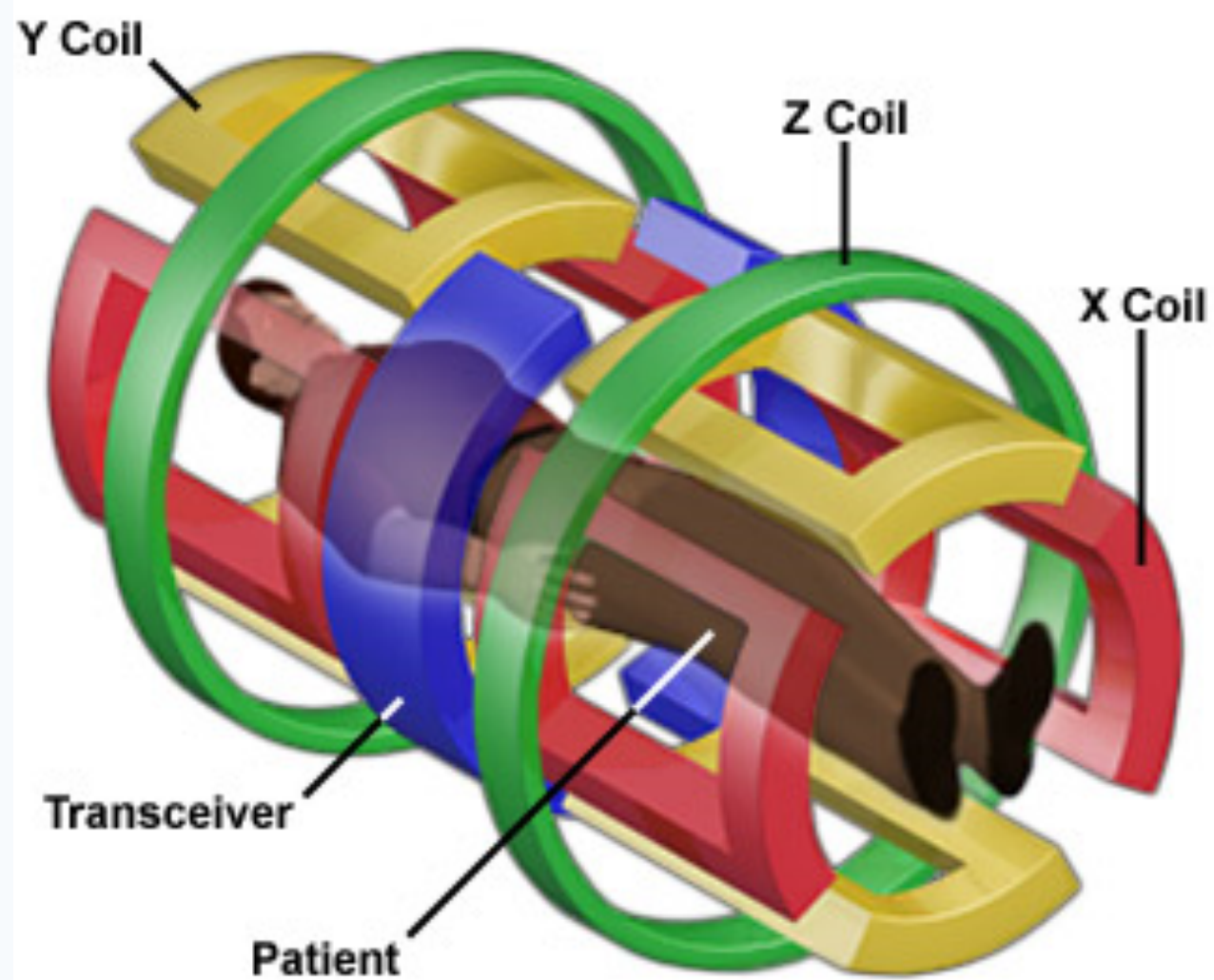
Receiver coils

Magnetic field gradient
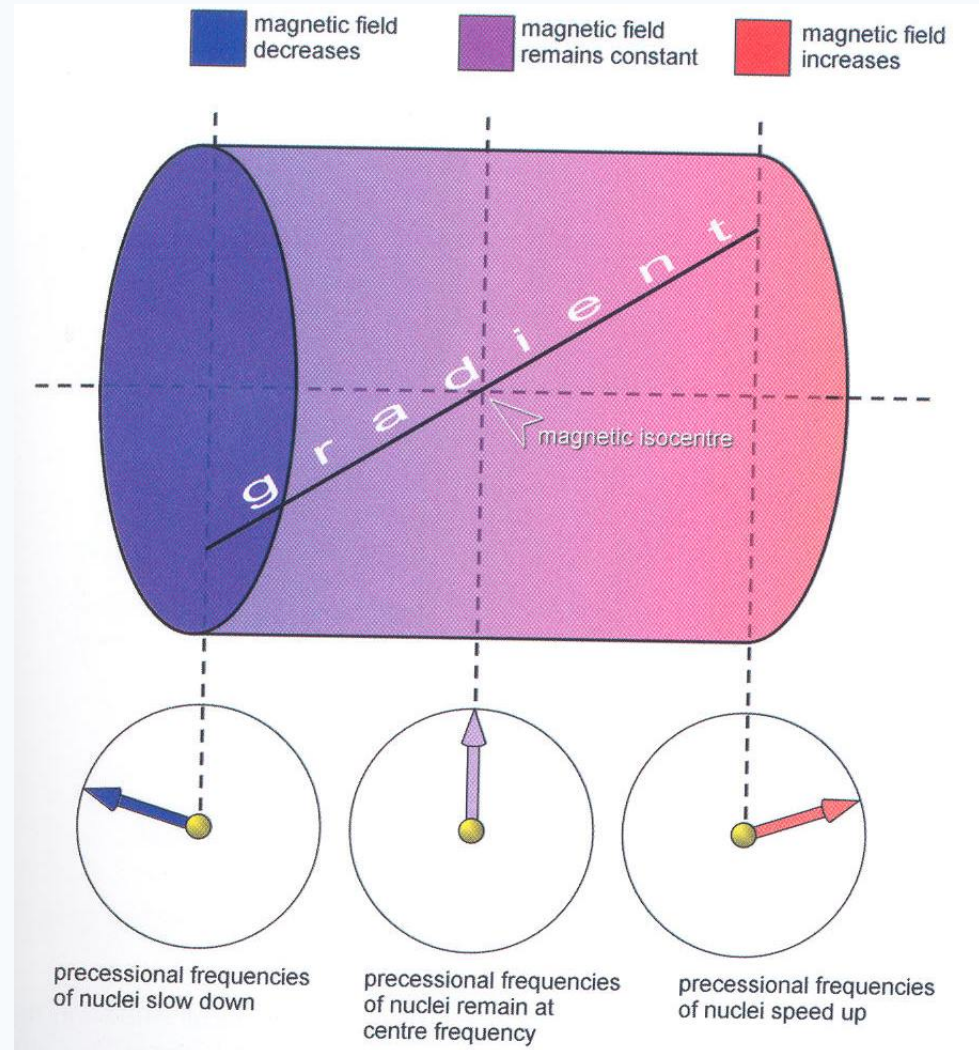
Effect of gradient
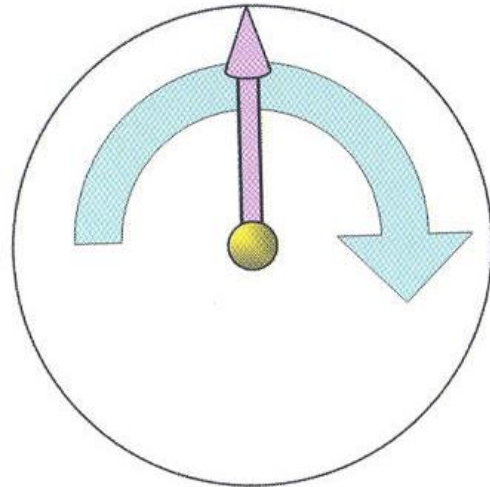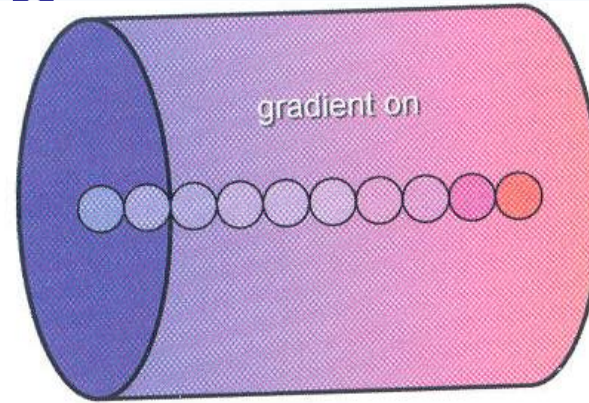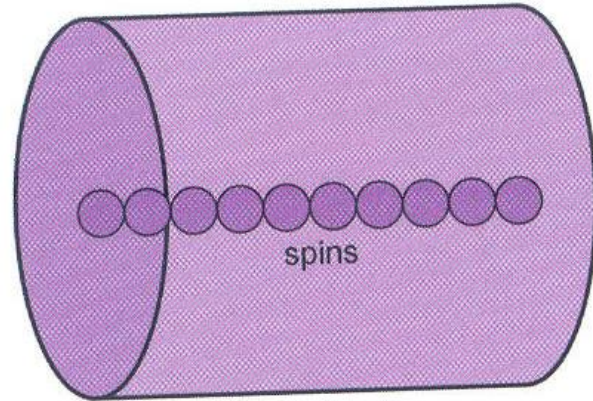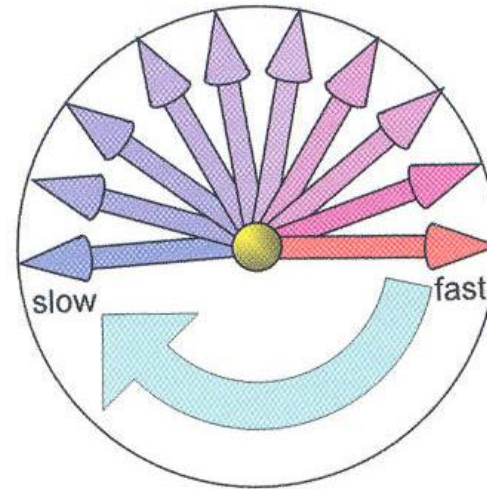
Gradient echo

Direction of the gradient switches
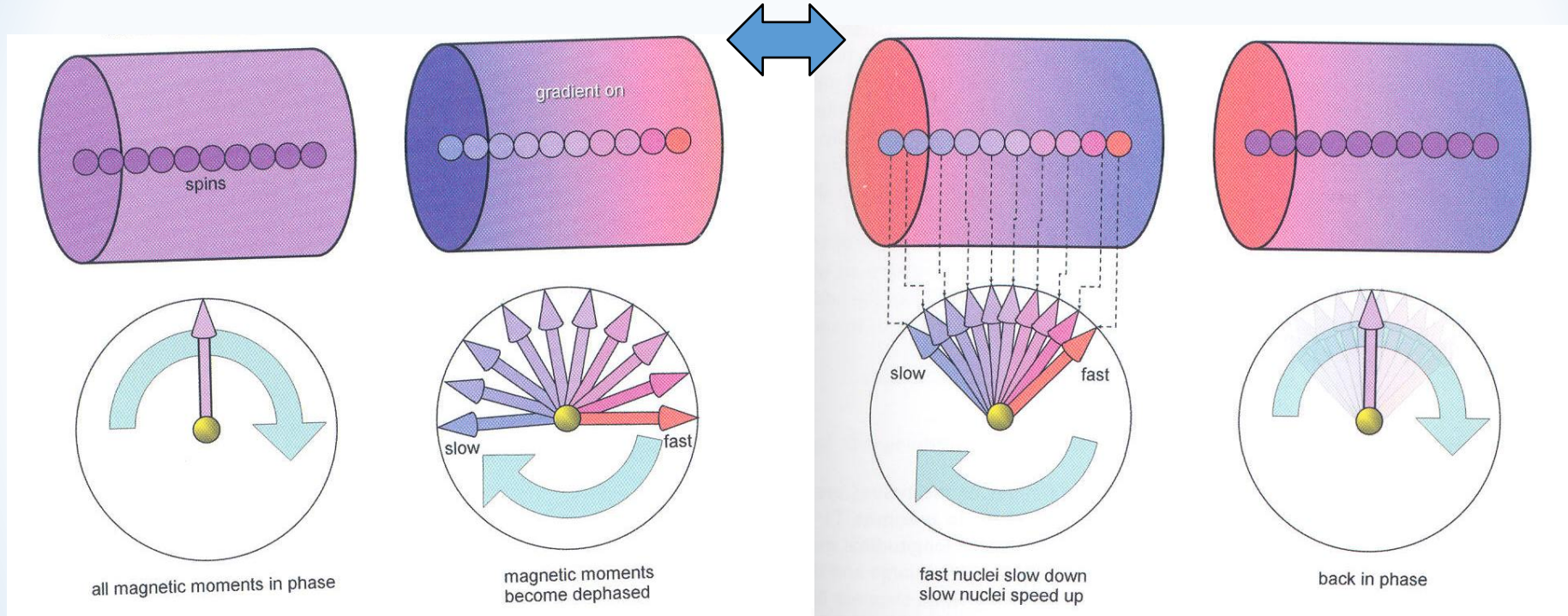
Signal from gradient echo is much weaker than one from spin echo:
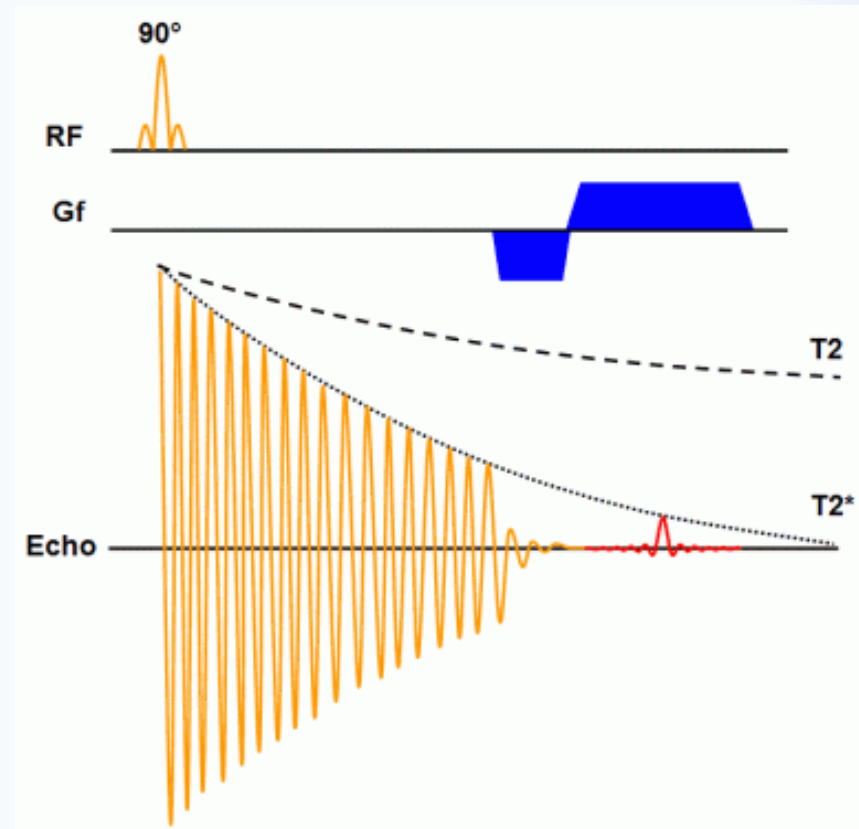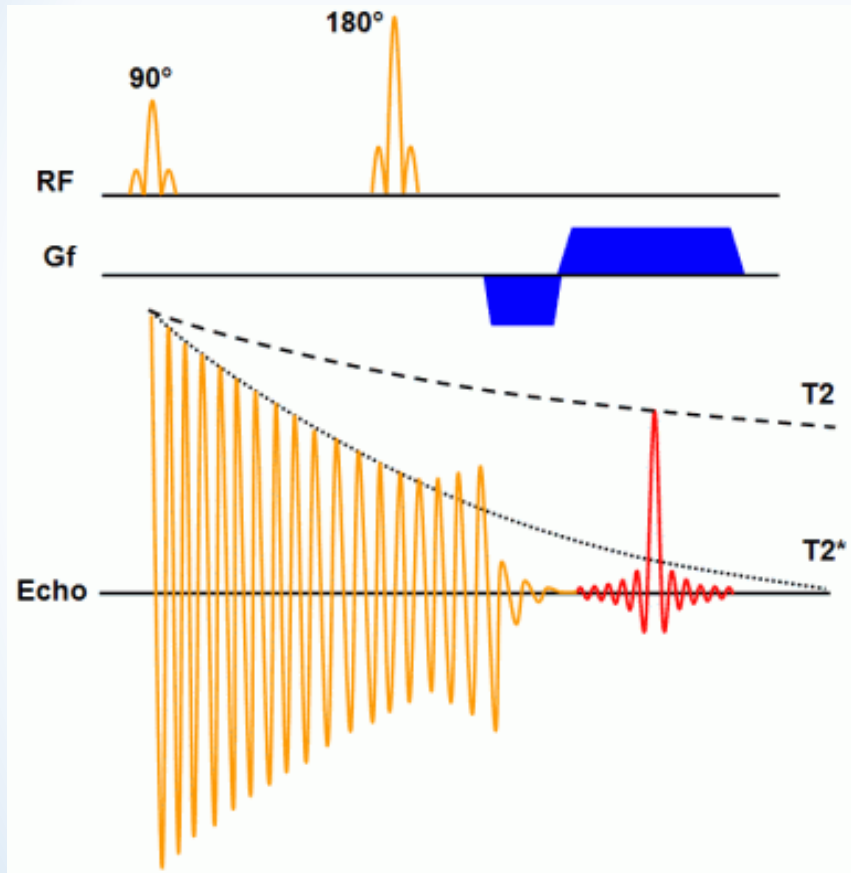
However, gradient echo sequences can be really fast,
when short TR and TE times are used.

Actual imaging sequence requires gradients anyway. We will take an spin-echo sequence as an example.
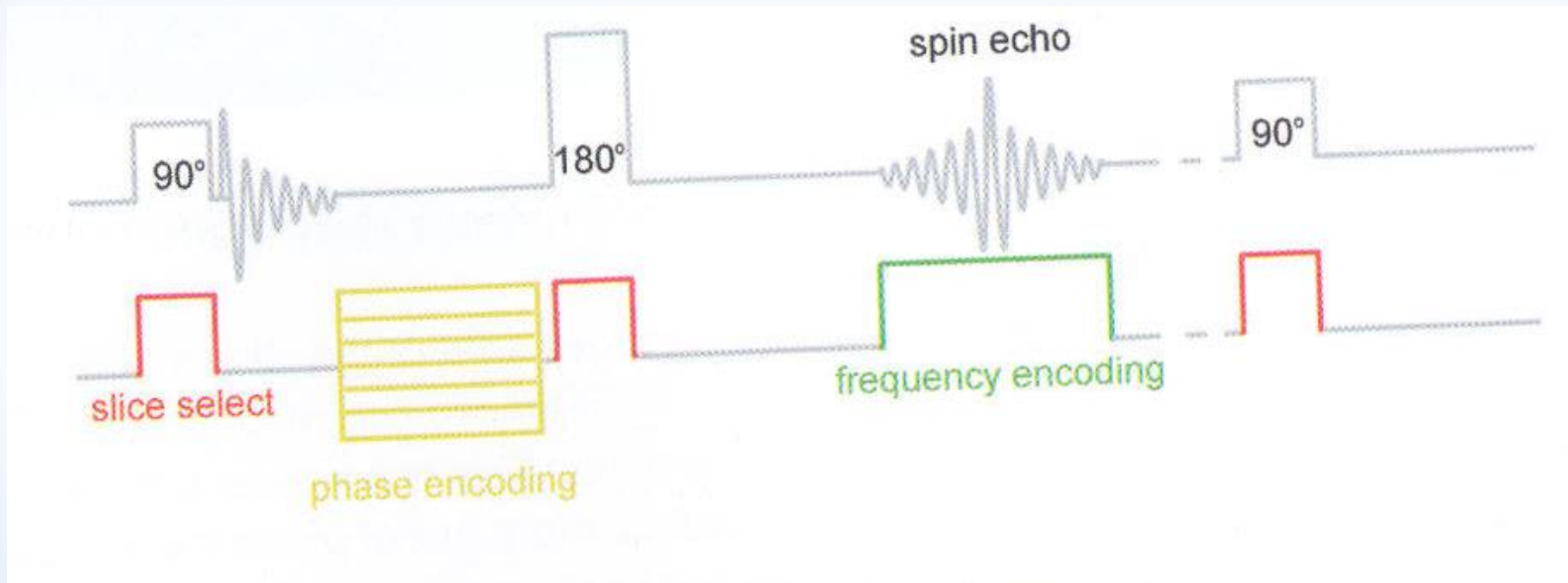
Effect of gradient
(remember Larmor frequency)

RF at 41.20 MHz resonates the spins at slice position A

RF at 43.80 MHz resonates the spins at slice position B
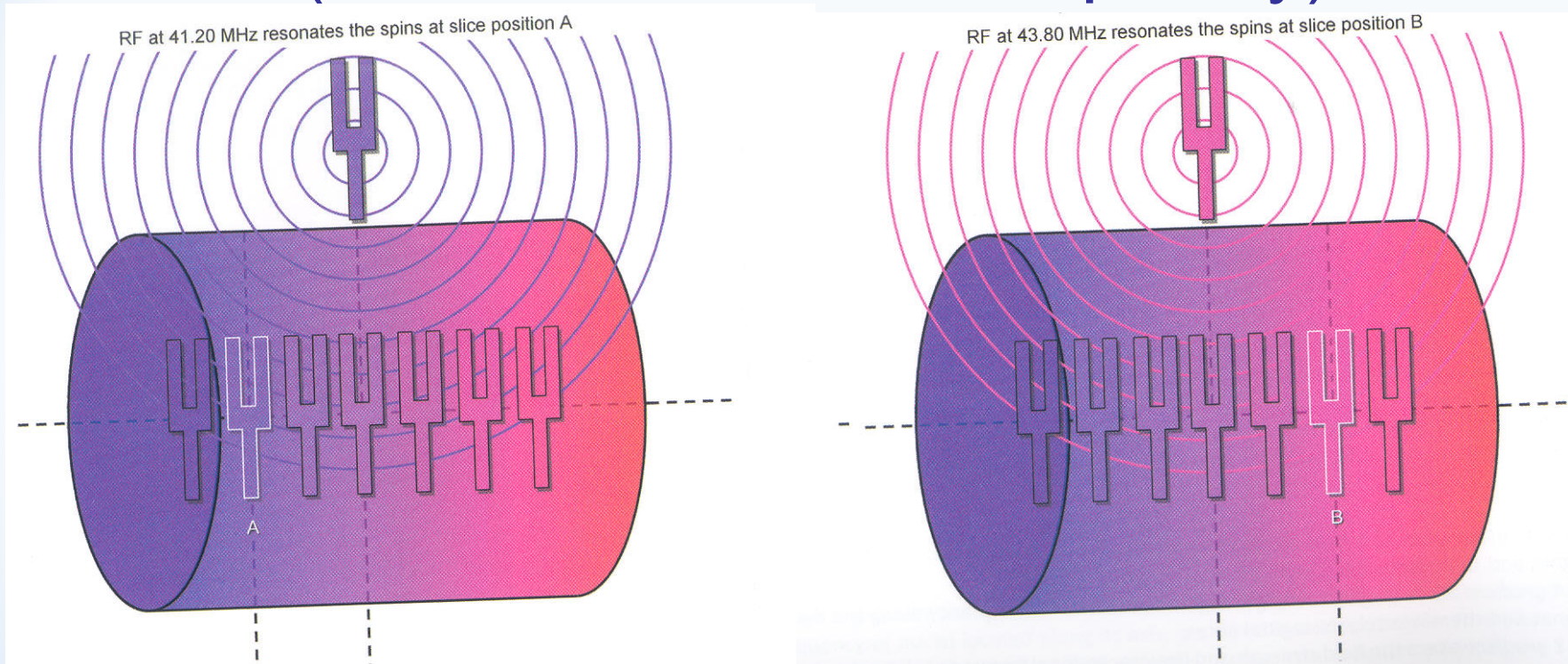
Localizing spins

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
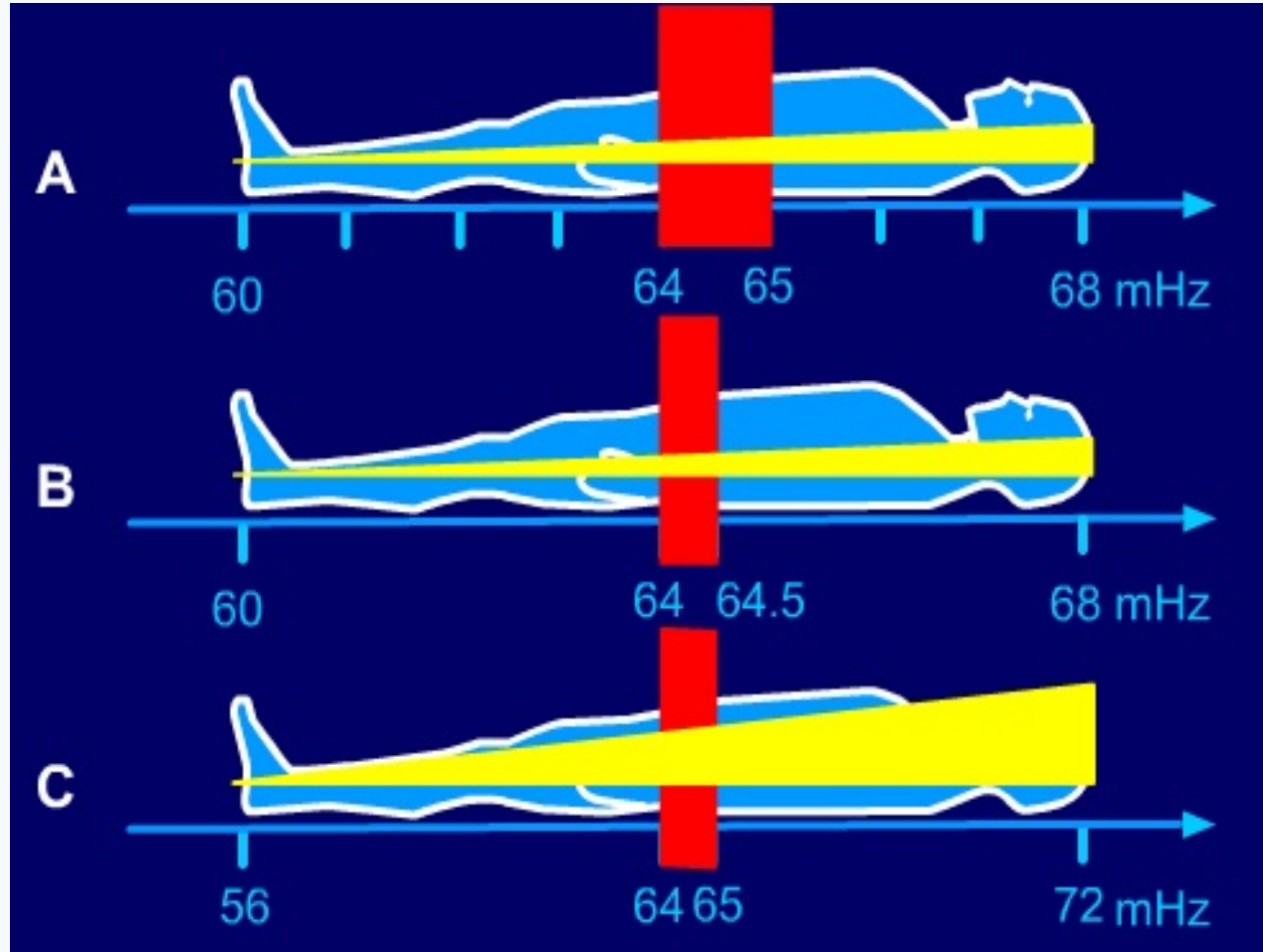
RF pulse shape
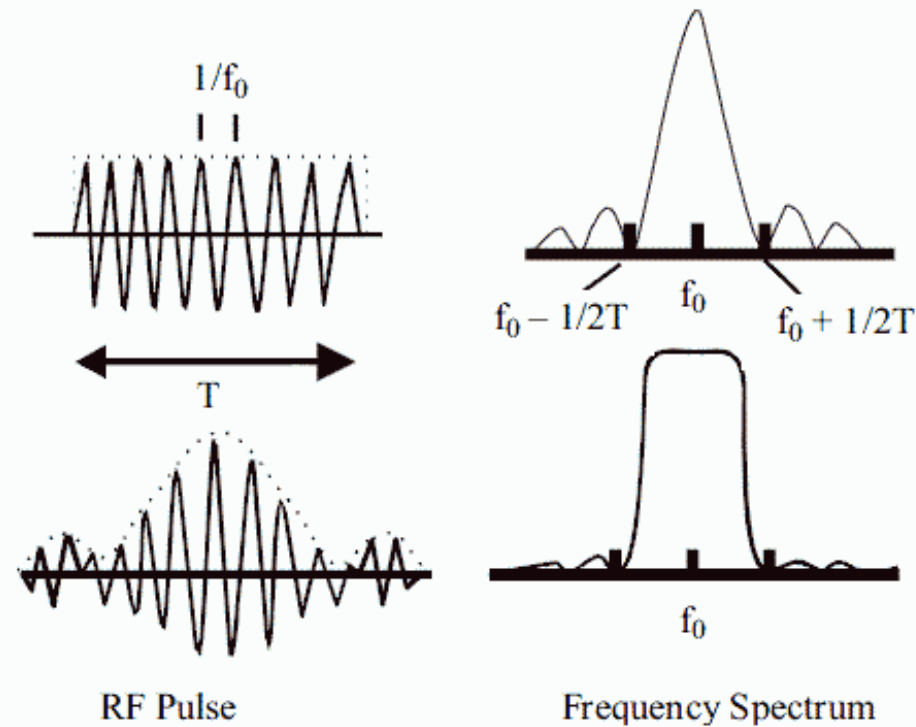
Figure 6.9 A pulse of pure sine waves lasting T seconds gives rise to a sinc function $(\sin(f)/f)$ frequency spectrum. The main central lobe of this spectrum occupies a range of frequencies, $\Delta f \sim 1/T$. This provides an approximate definition of the bandwidth and, in MRI, the slice width. There 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
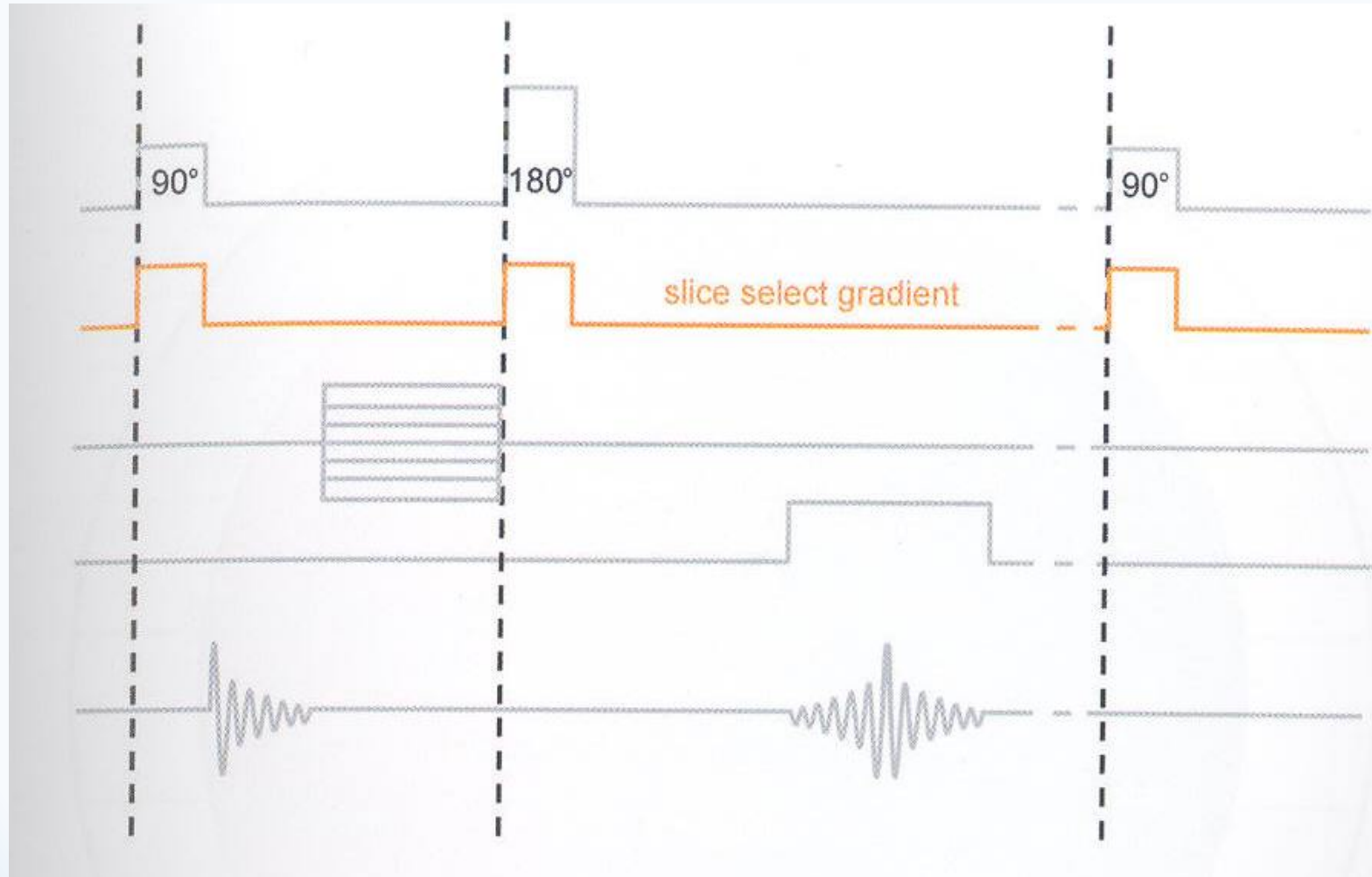
Phase encoding gradient
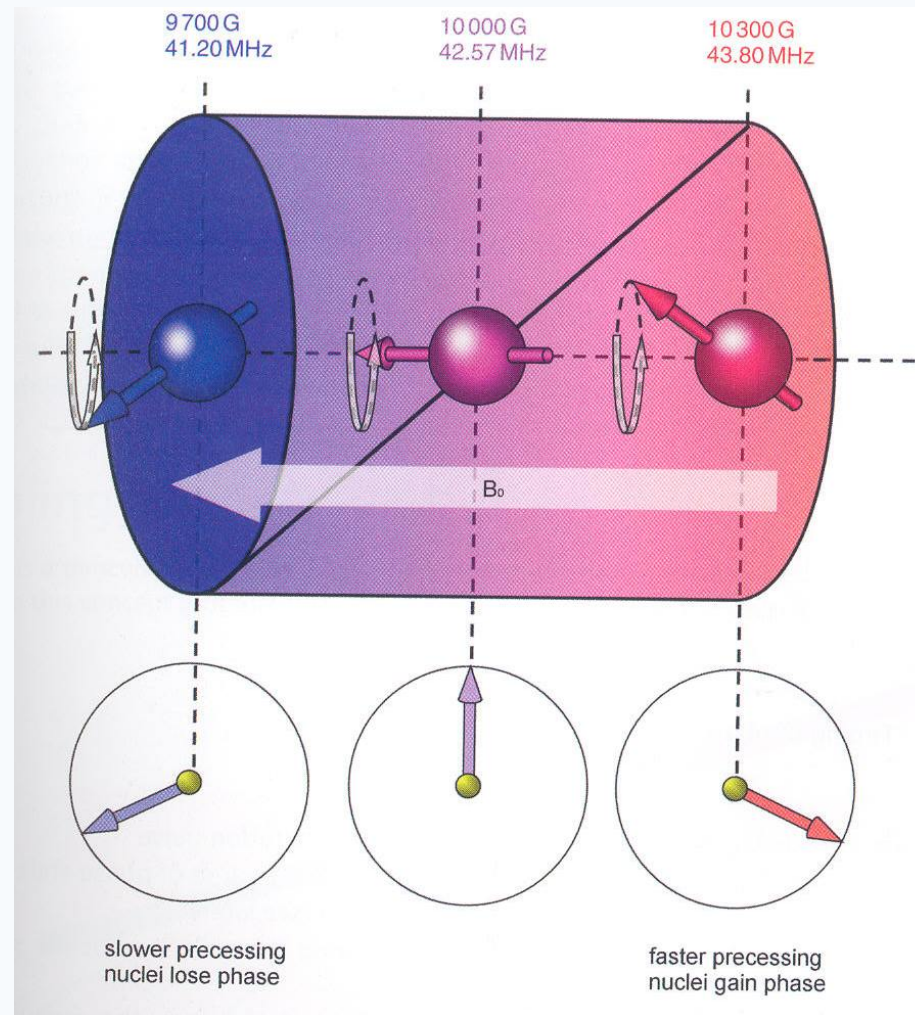
Phase encoding gradient
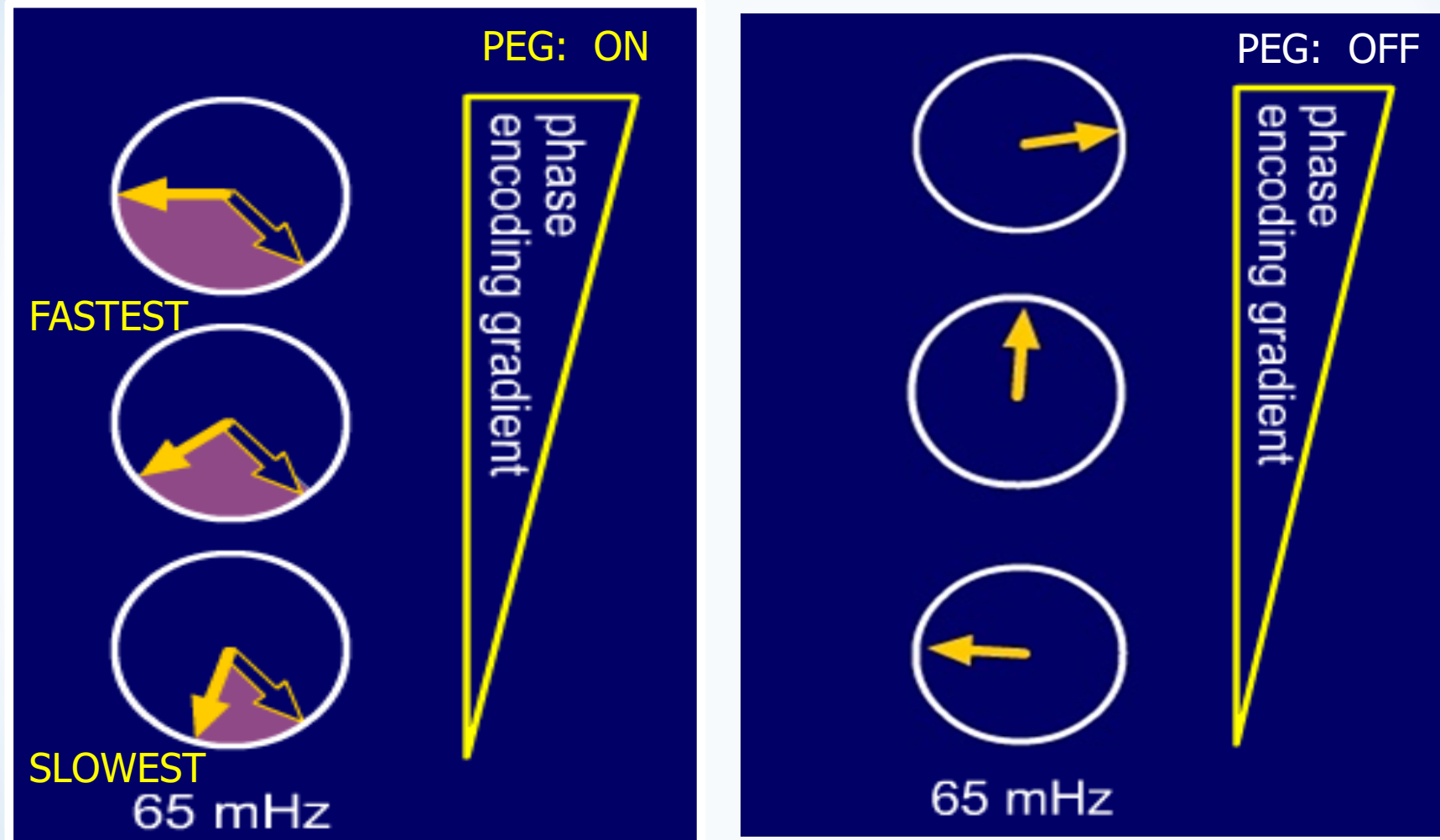
After PEG is switched of spins are localized by their phase difference!

Phase encoding gradient
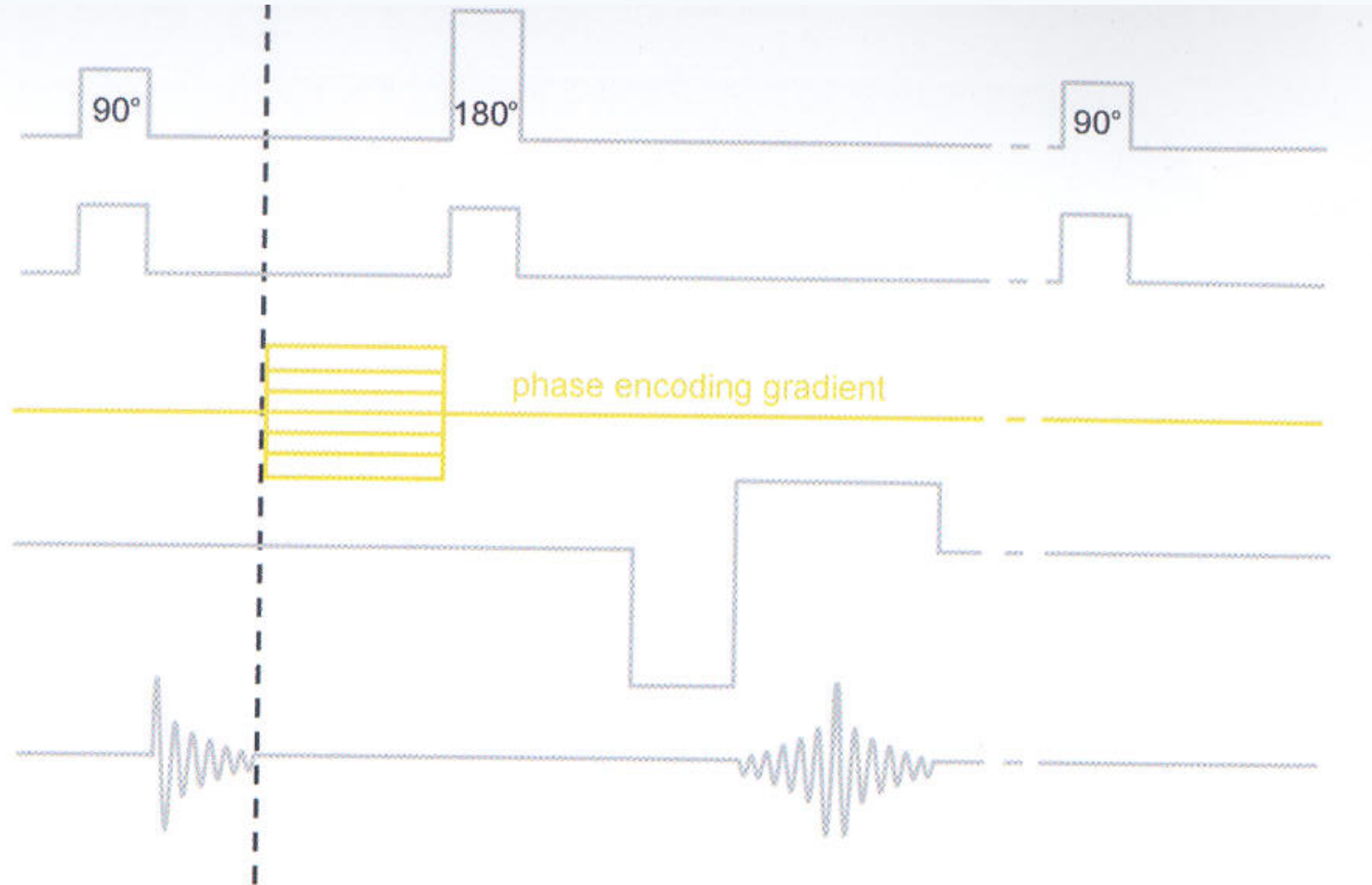
Frequency encoding gradient
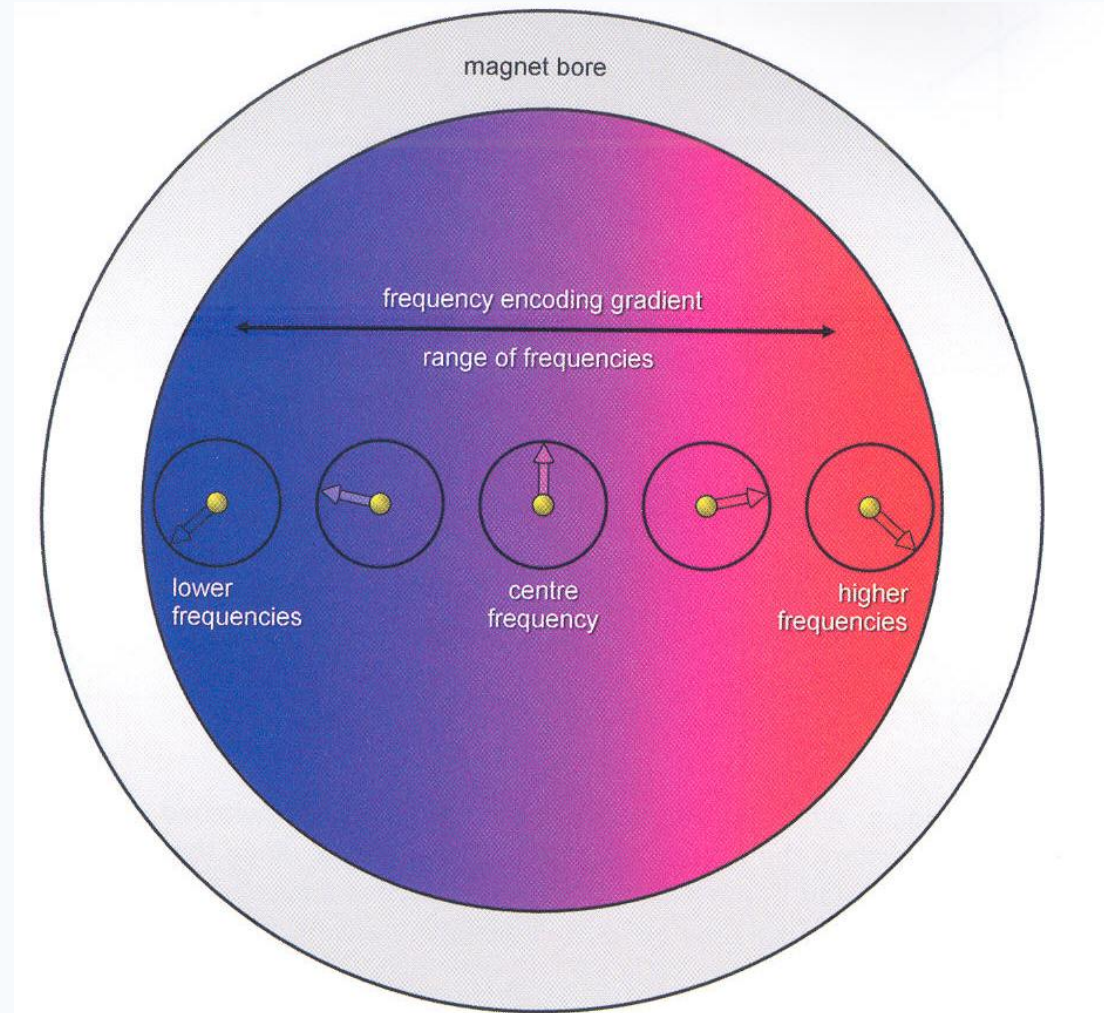
Frequency encoding gradient

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

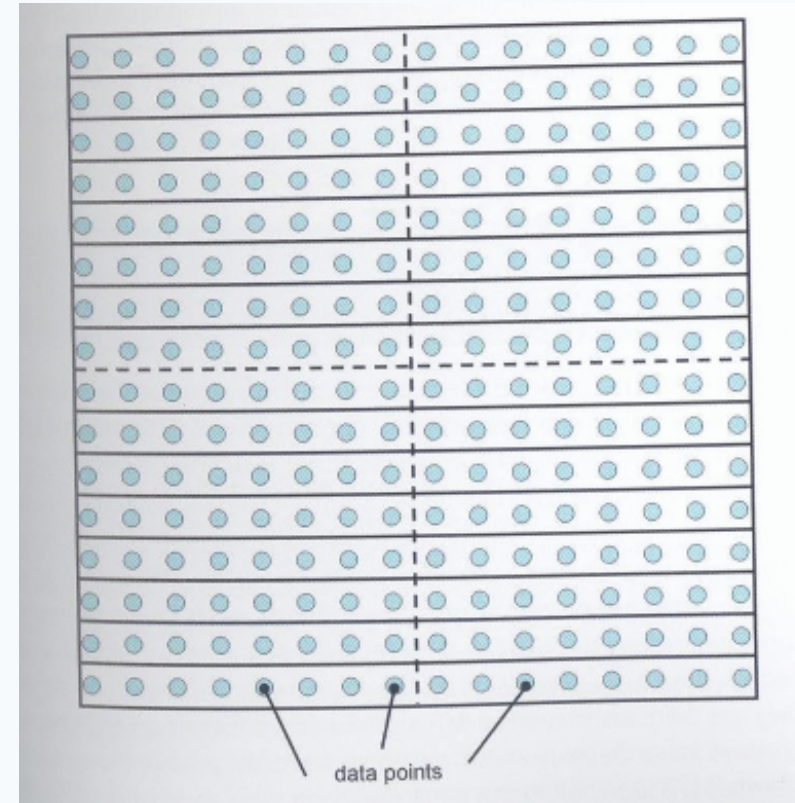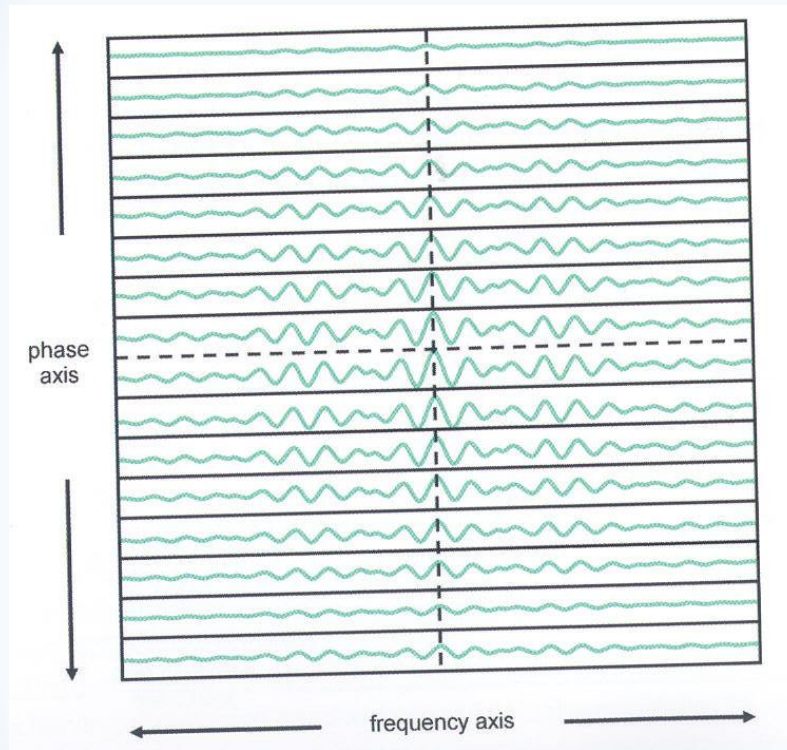Different frequency during acquisition gives information about the location of the spins.
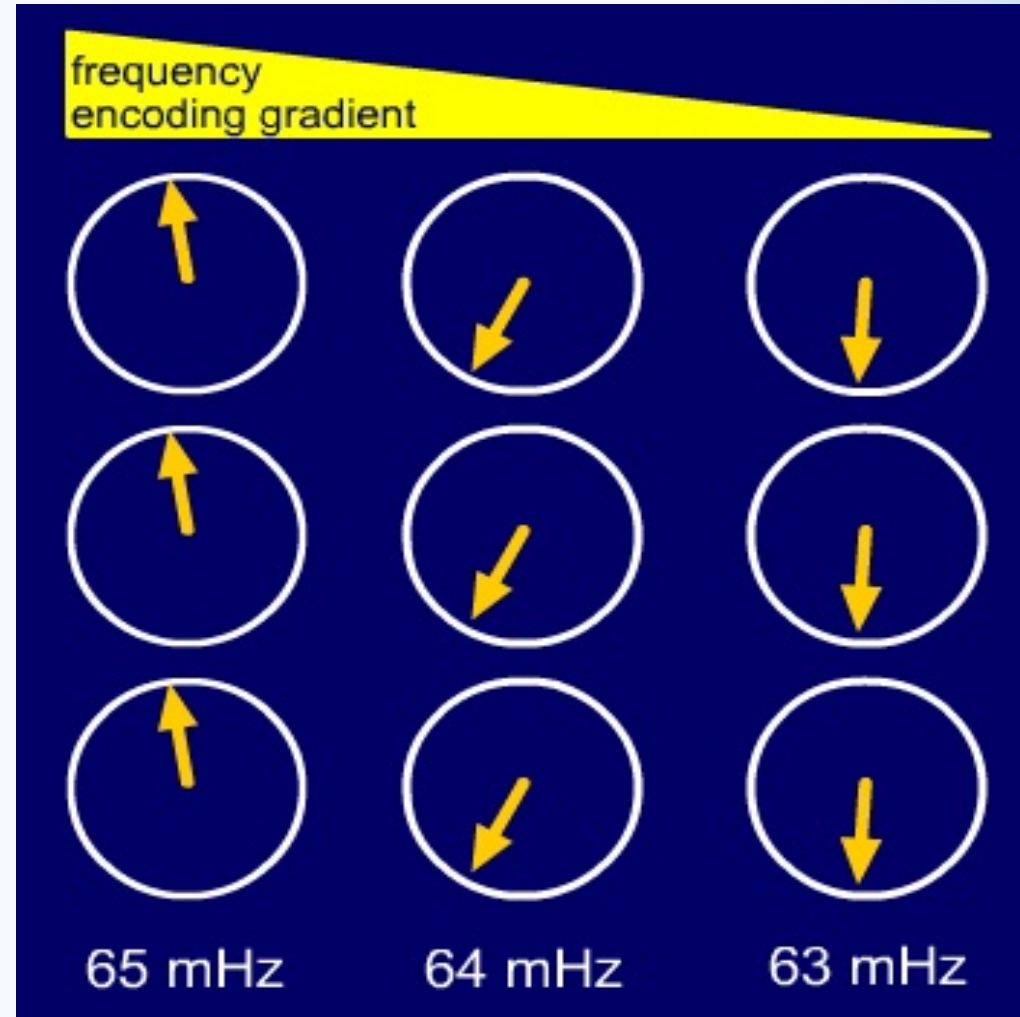
Frequency encoding gradient
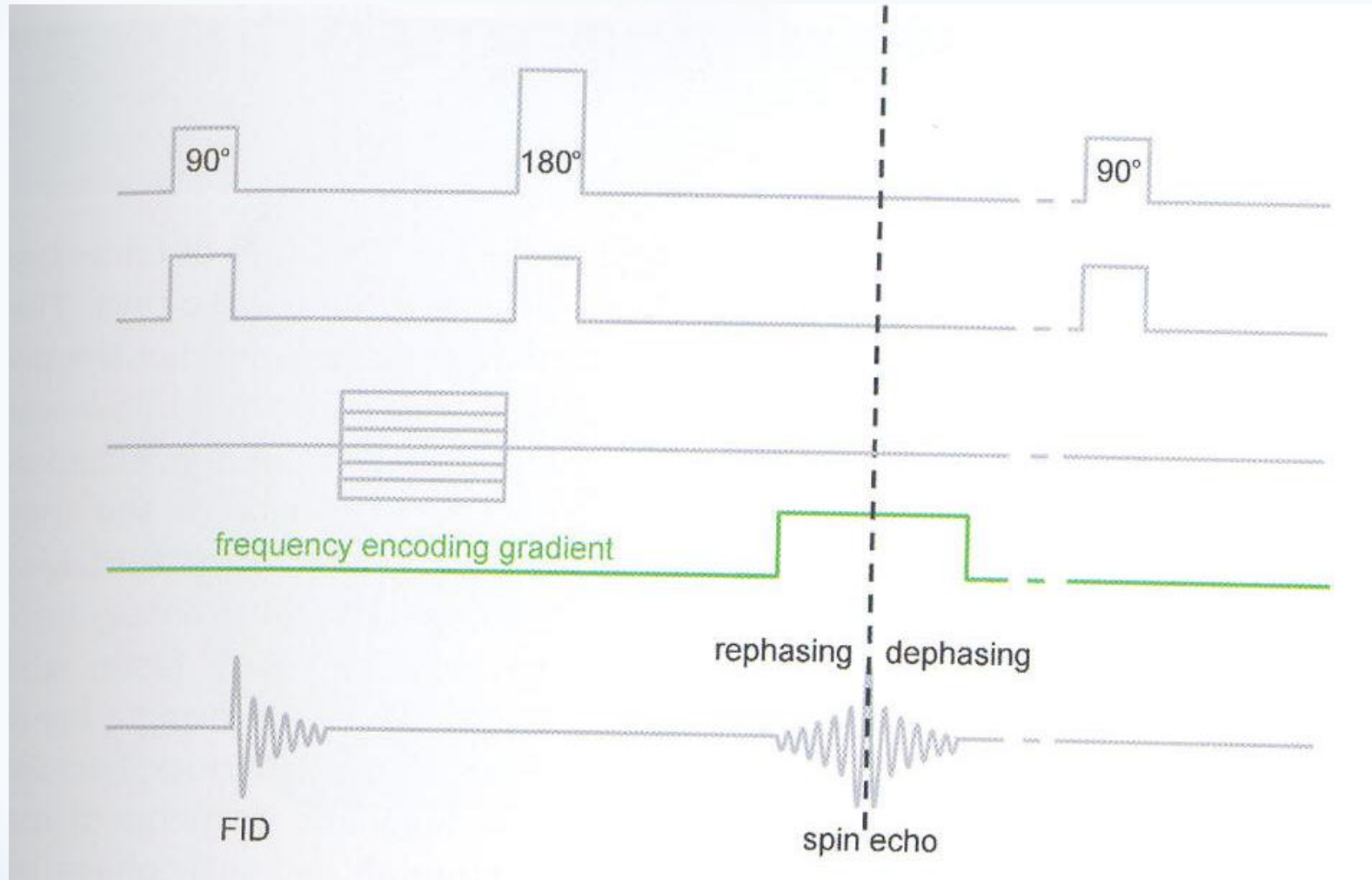
Spin echo gradient and RF- pulse diagram

GRADIENTS

G_z Slice Selection

G_y Phase Encoding

N_y phase encodings

G_x Frequency Encoding

RF PULSES

N_x data points

RF SIGNAL

2D FT- MR imaging sequence

time t
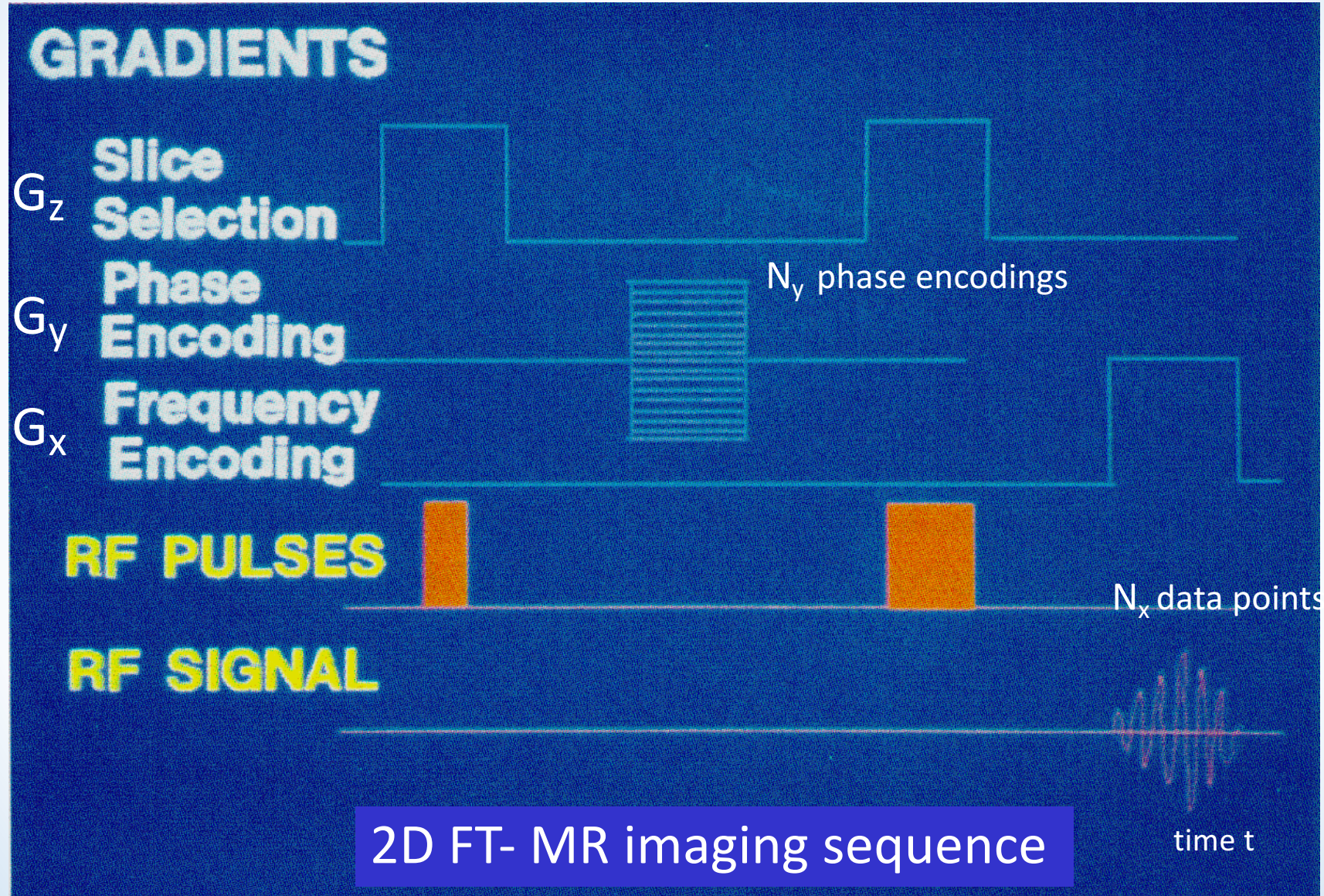
K-space / data points (imaging matrix)

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
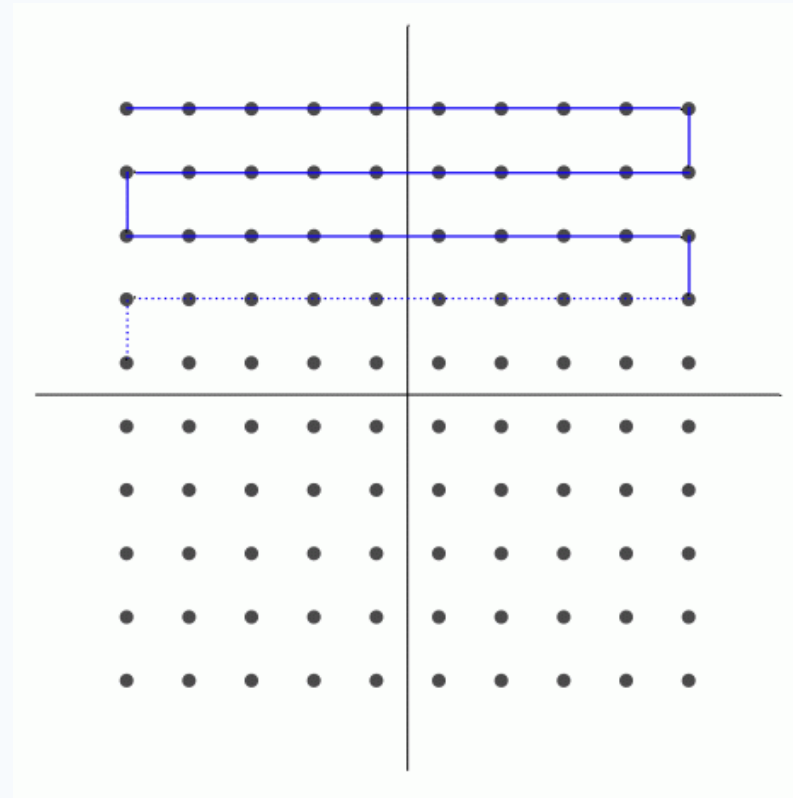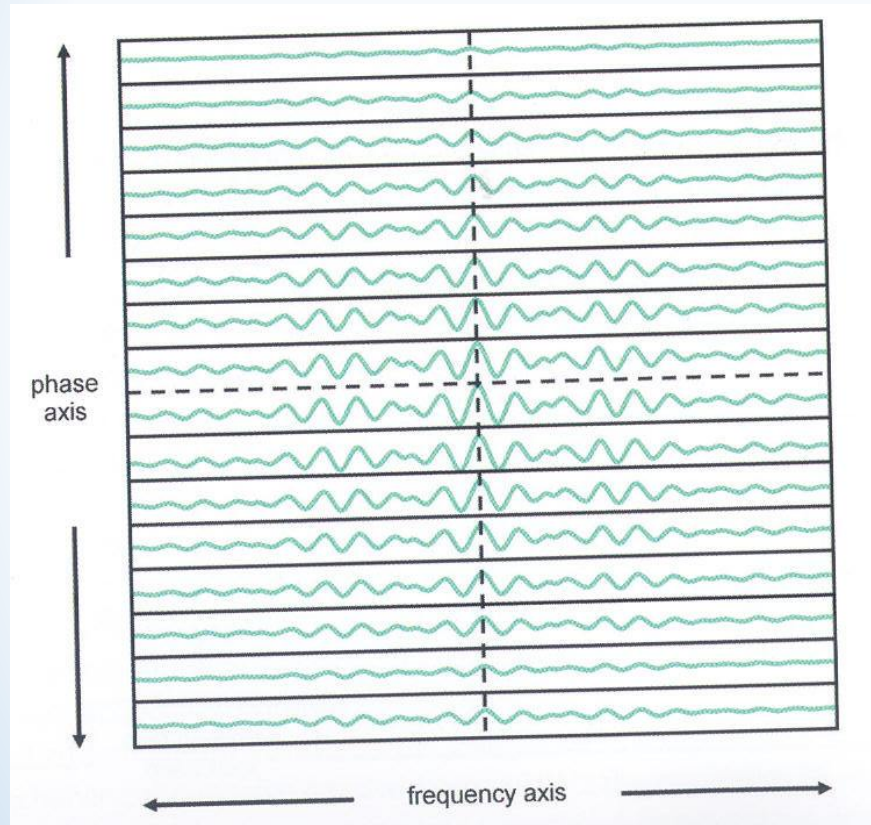
Gradient echo sequence contains exactly same gradients, only 180 degree pulse and it's slice encoding gradient is missing.
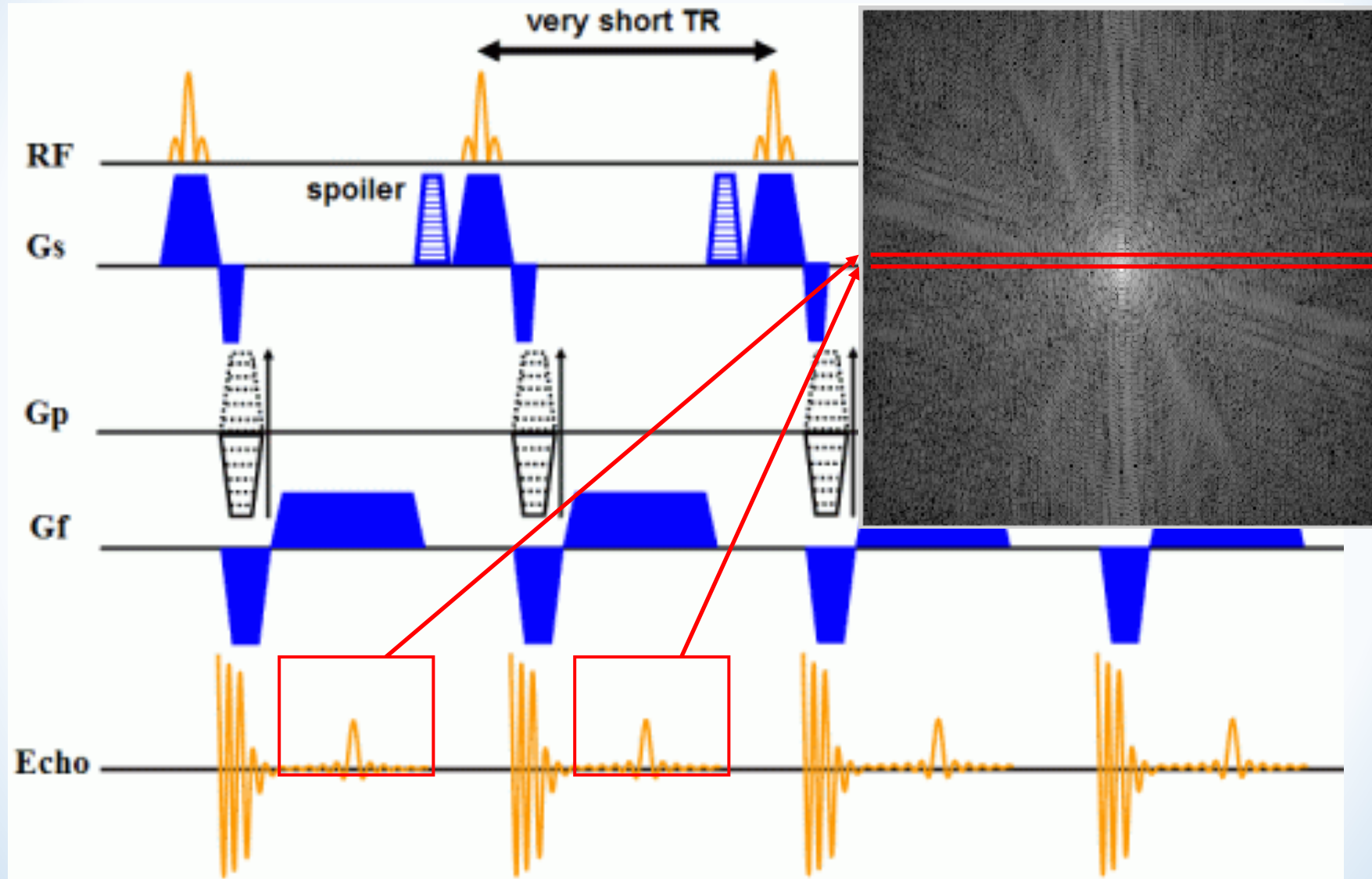
In conclusion, gradients are used for localization of the acquired signal.

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
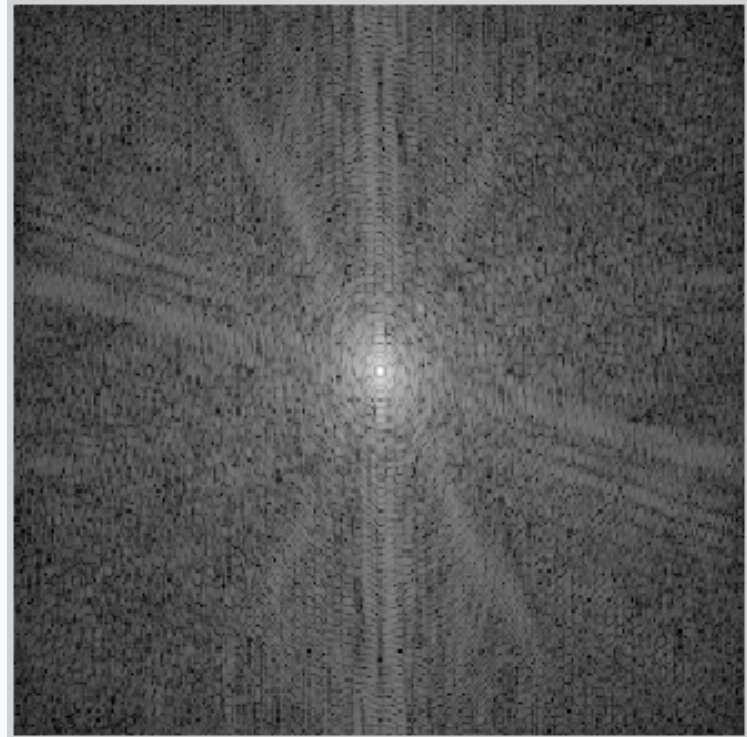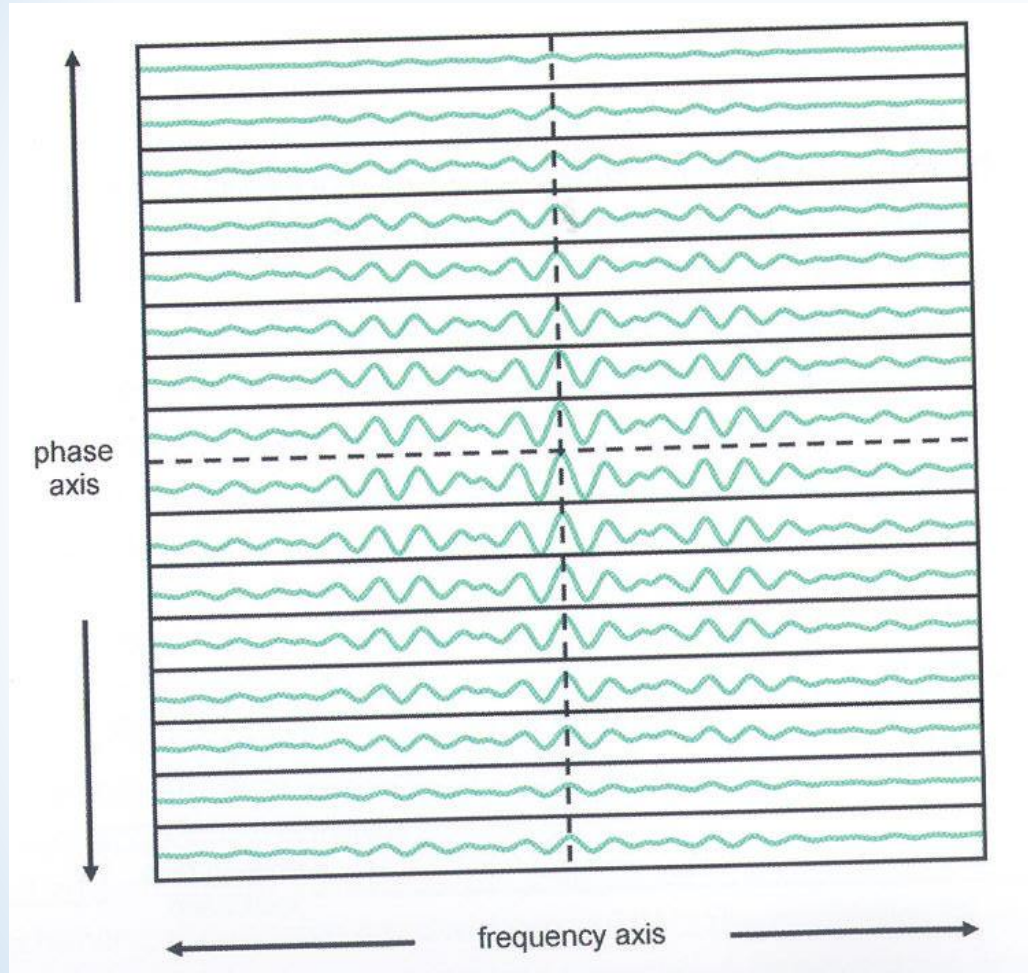
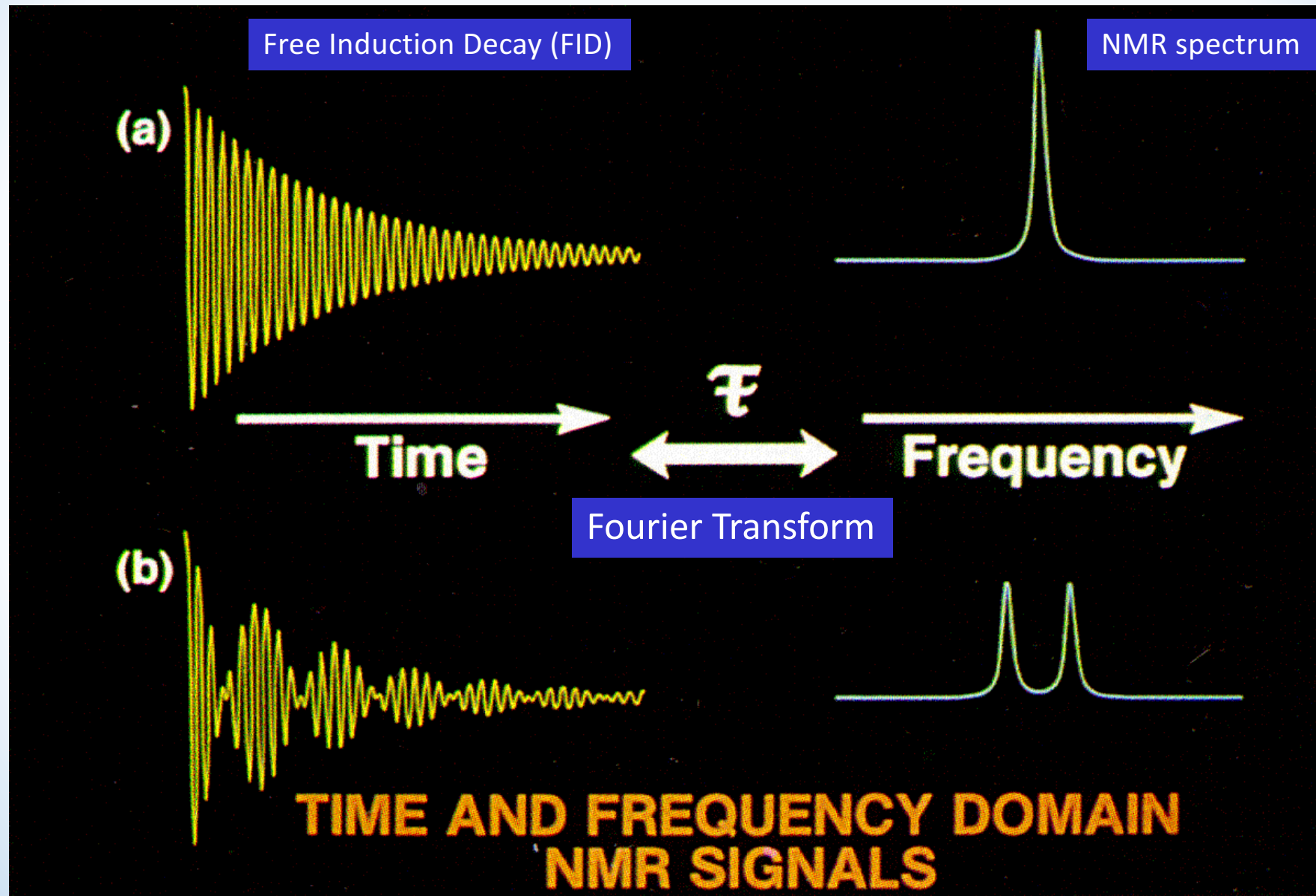**By applying 2D Fast Fourier Transformation, we can get an image from the acquired data of k-space.**
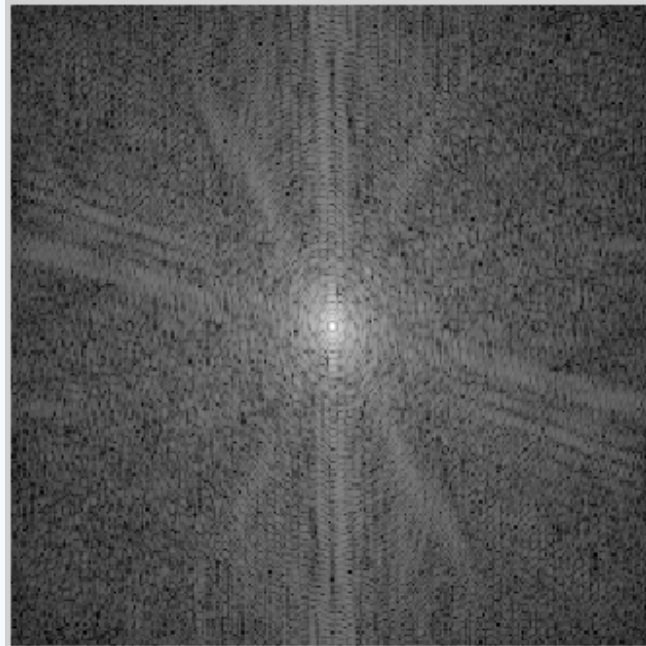
2D FFT
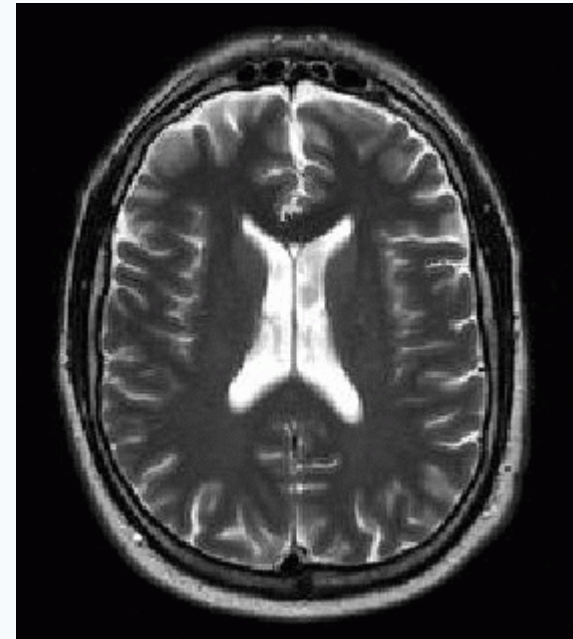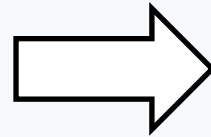
Data in phase encoding direction is in frequency domain as well. It is just collected using different approach.

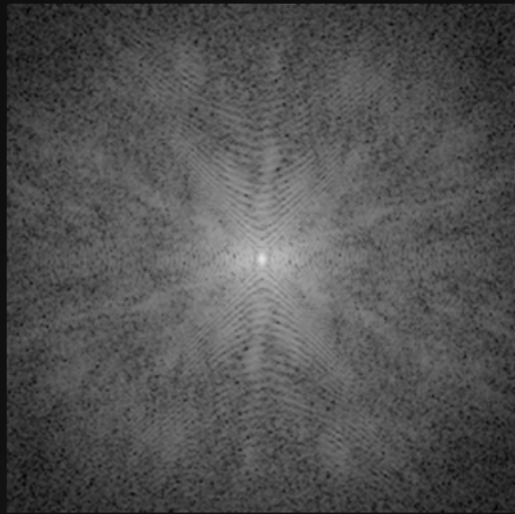# Fourier Transform of FID = NMR spectrum

2D- MR image reconstruction

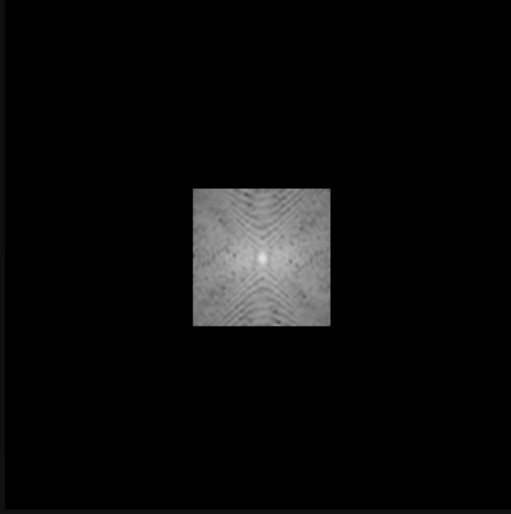Raw data matrix = k-space
Matrix size 256x256

Image reconstruction from k-space to "image space" using 2D Fourier transform
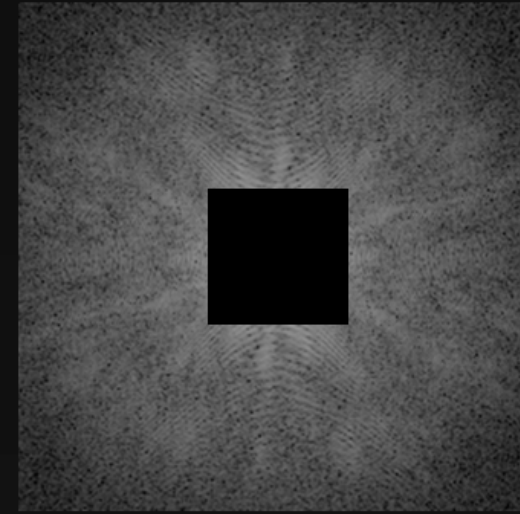
(2DFT)
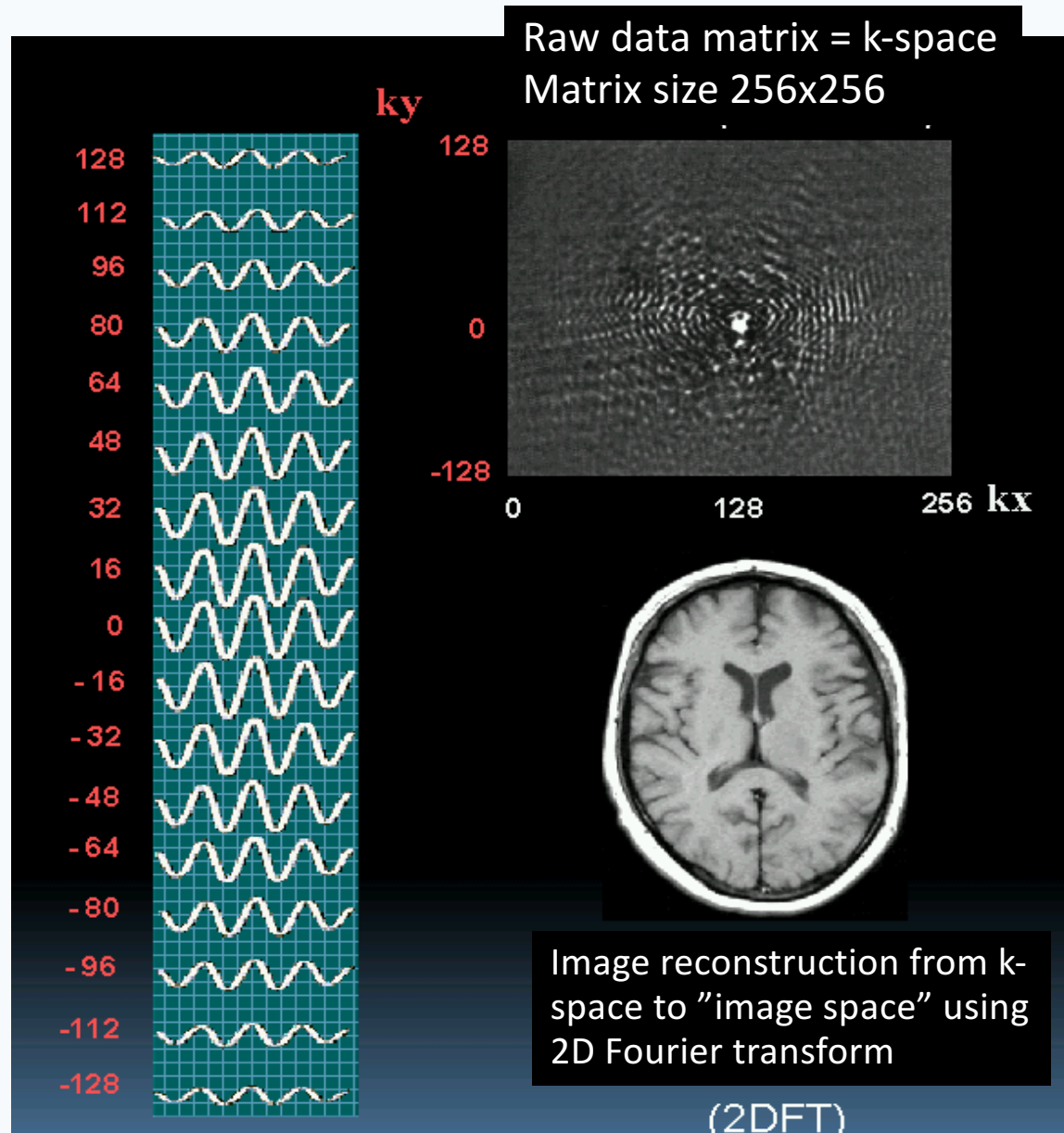
Full Image

Intensity-Heavy Image
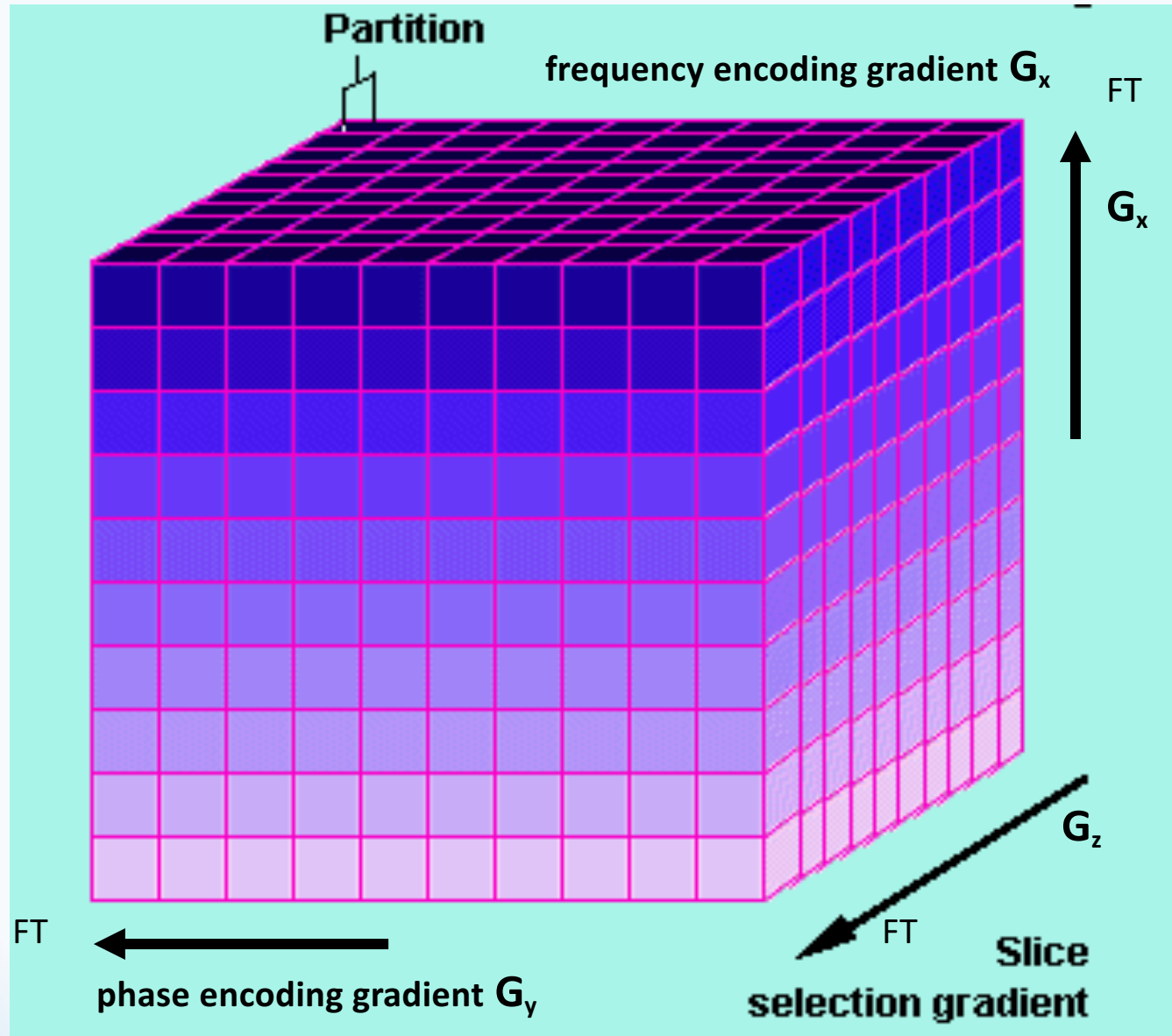
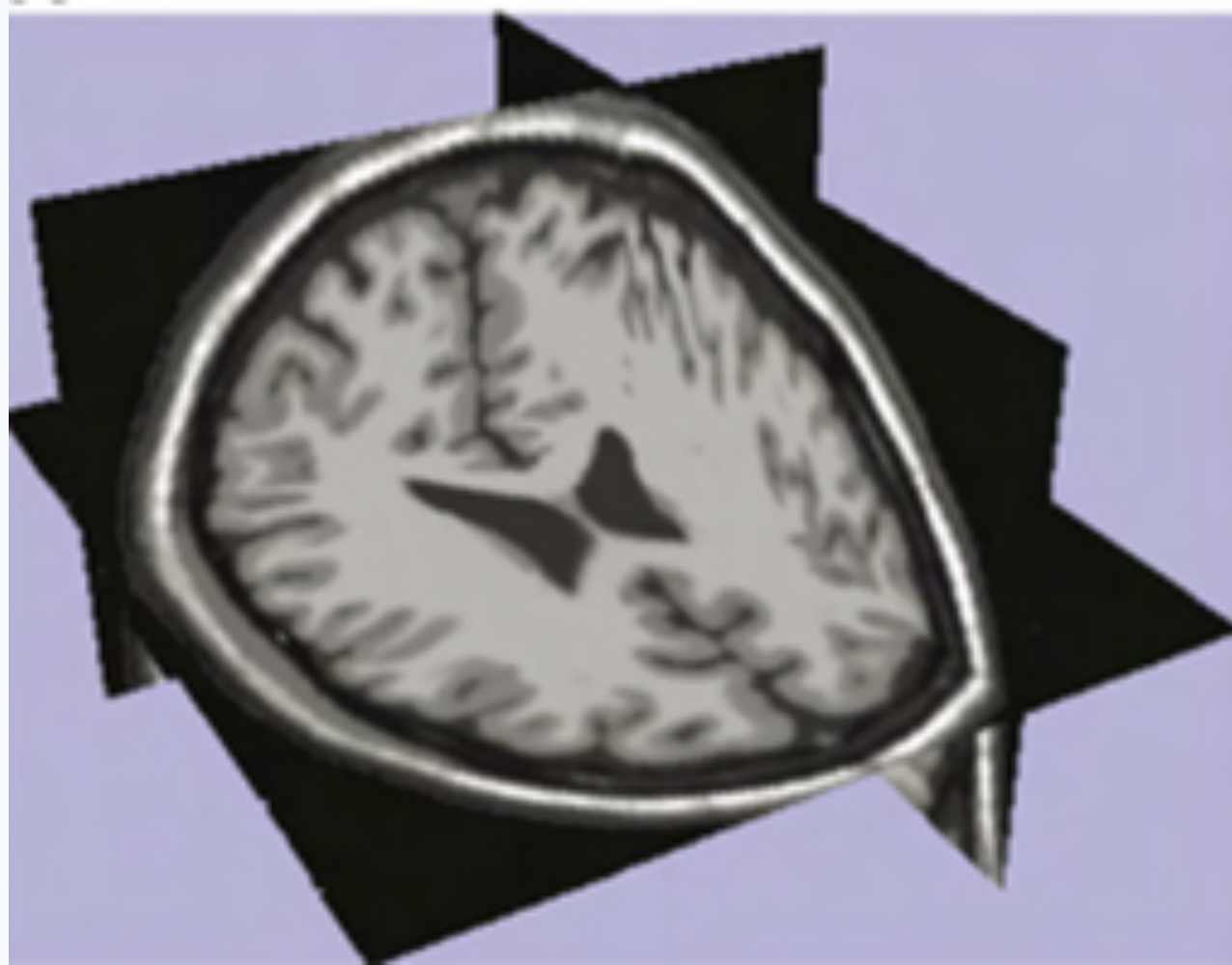Detail-Heavy Image

Full k-space

Lower k-space

Higher k-space
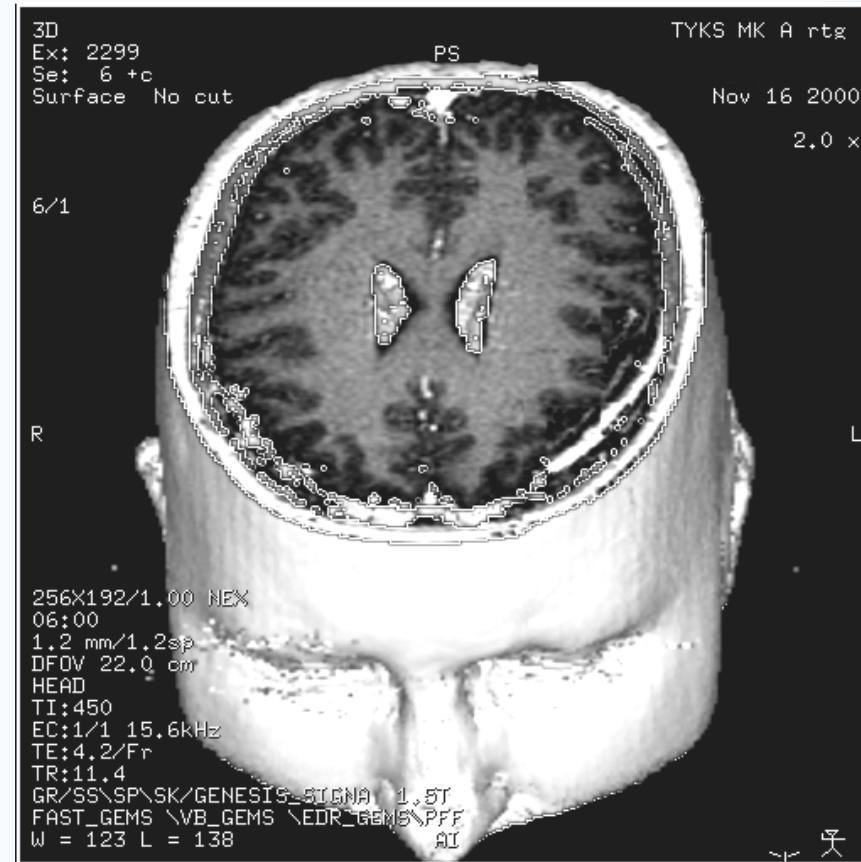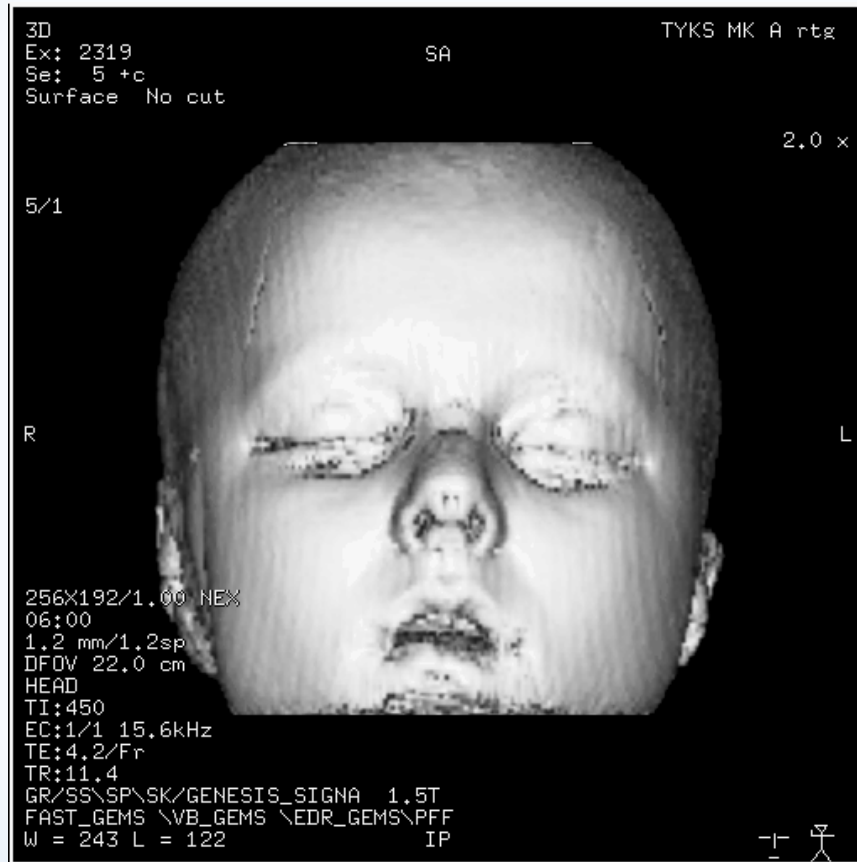
3D- FT MR image reconstruction

3D- reconstruction of head MR image

voxel size = 0.86x0.86x1.2 mm^3

3D- reconstruction of head MR image

$N_X \times N_y \times N_z = 256 \times 192 \times 124 = 6094848$ voxels

voxel size $= 0.86 \times 0.86 \times 1.2$ mm^3

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."
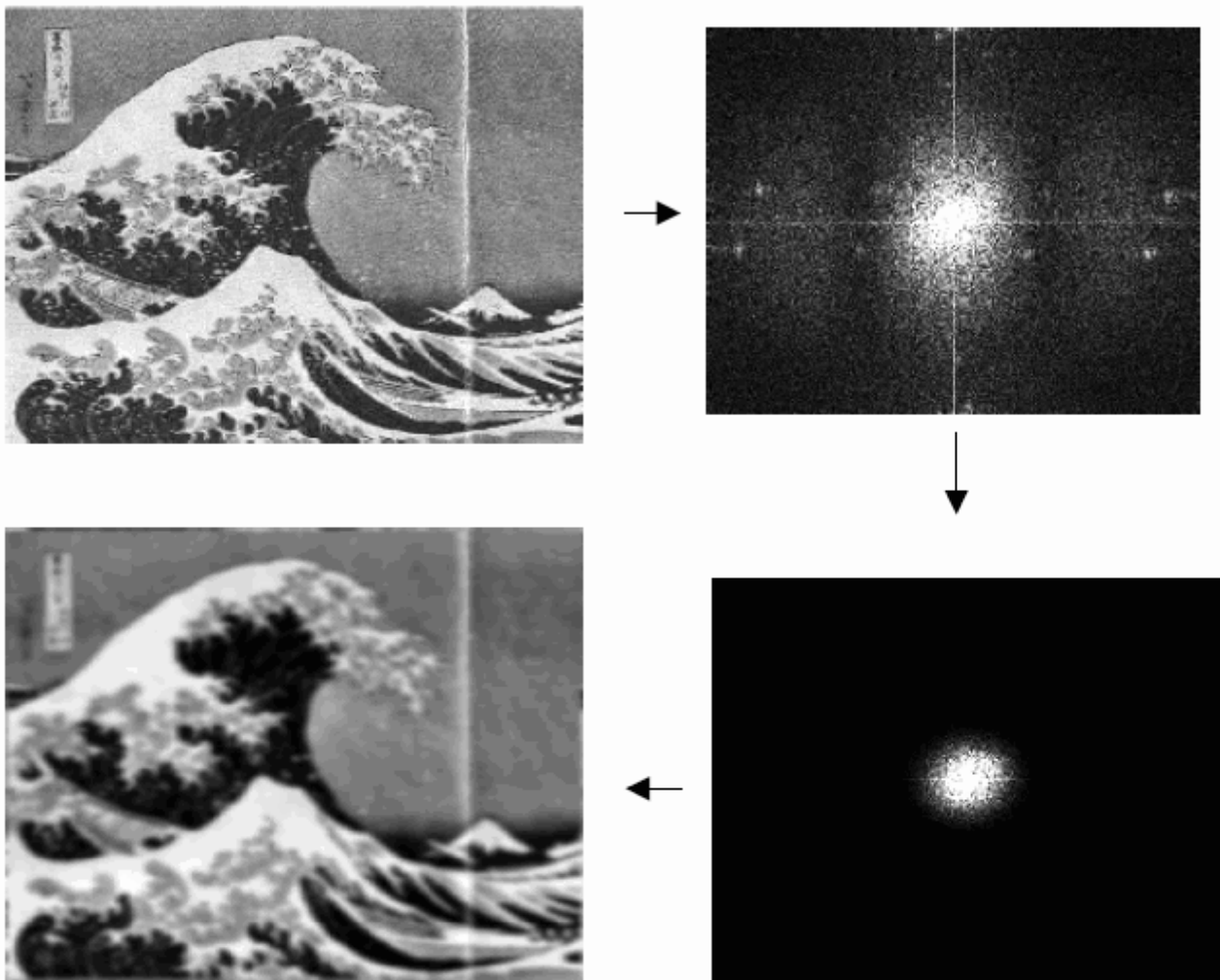
Figure 1.4 The monochrome Hokusai, in the top left panel, is Fourier transformed to produce its K-space map in the top right-hand panel. The K-space map is filtered, to reduce the amplitudes of the larger K values (shorter waves) in the bottom right-hand panel. The filtered K-space waves are then recombined 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.
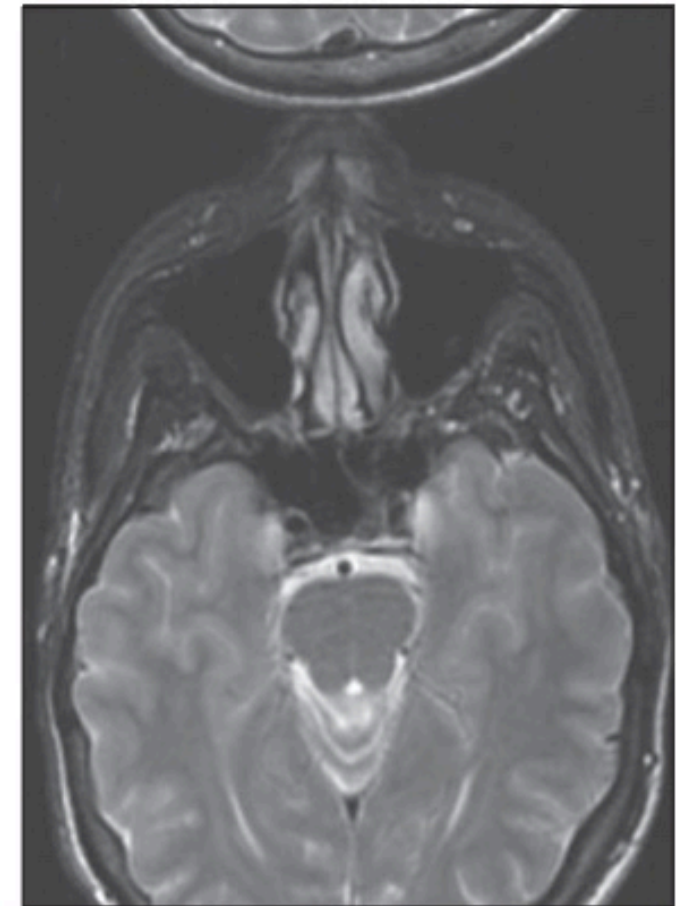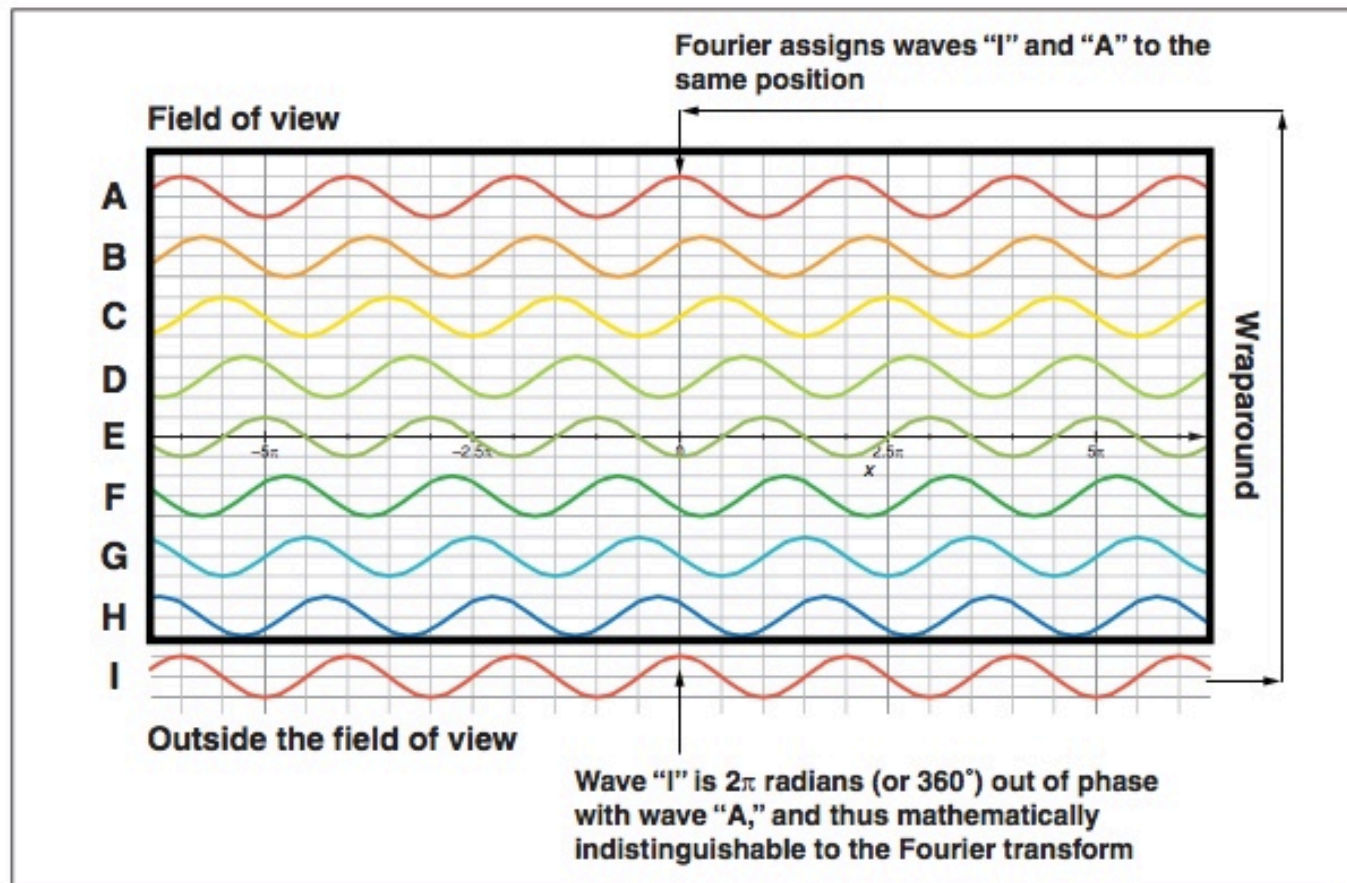
Fig. 12—Wraparound.
A, Wraparound (aliasing). Only phase shifts between 2π radians or 360° are available to encode an image. The phase shift in this image (depicted by waves A–H) covers the field of view (*black rectangular outline*). Just outside the field of view, wave "I" has assumed a phase shift 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.
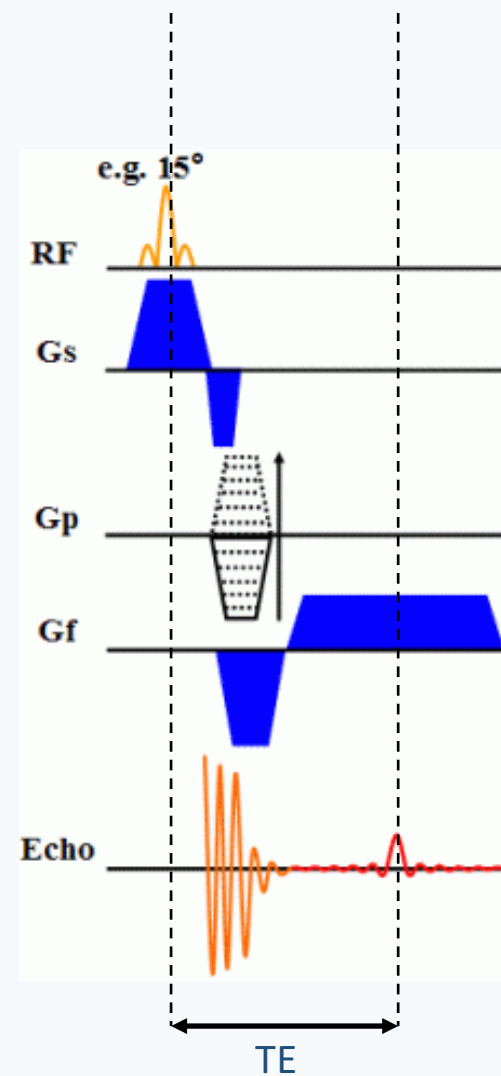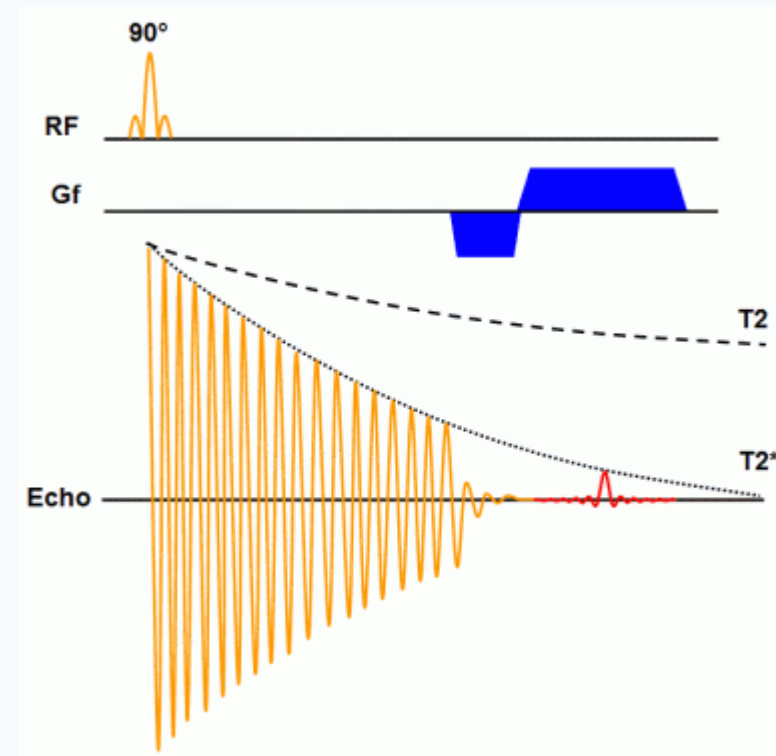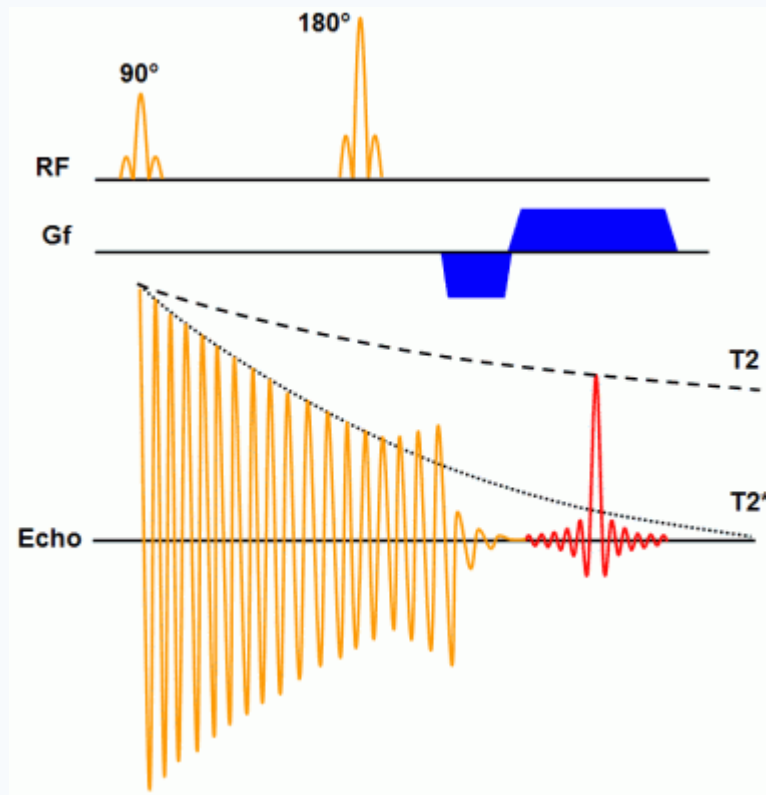
Gallagher et al AJR (2018) Intro to Fourier Transform

Zipper artefact

Gibbs ringing artefact
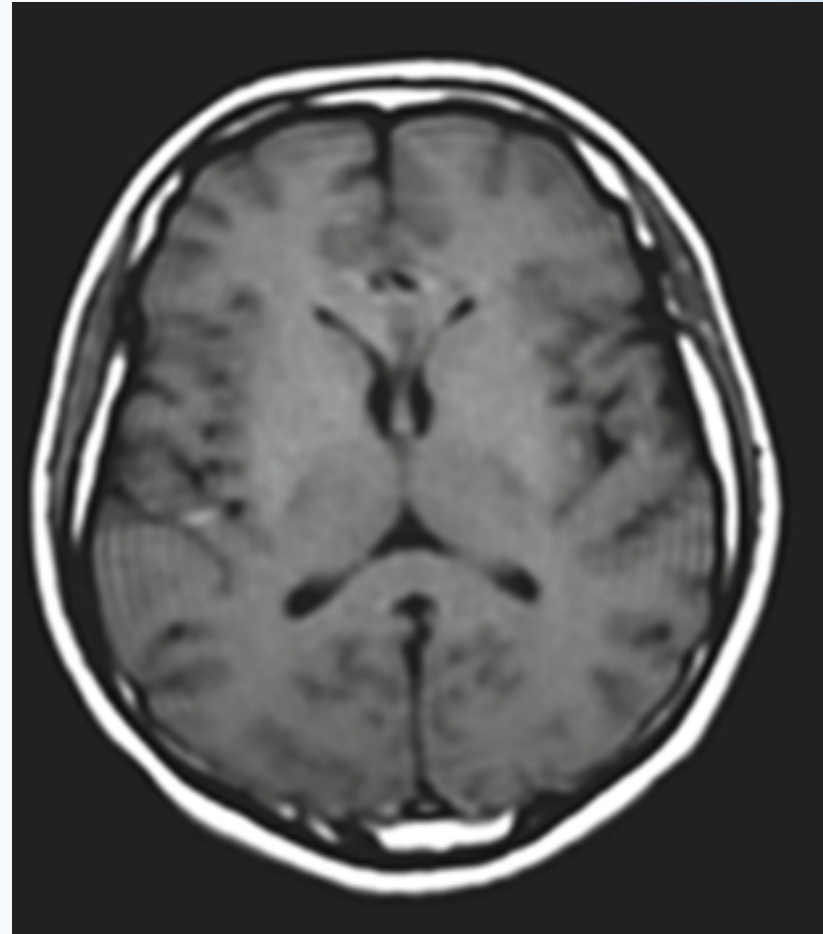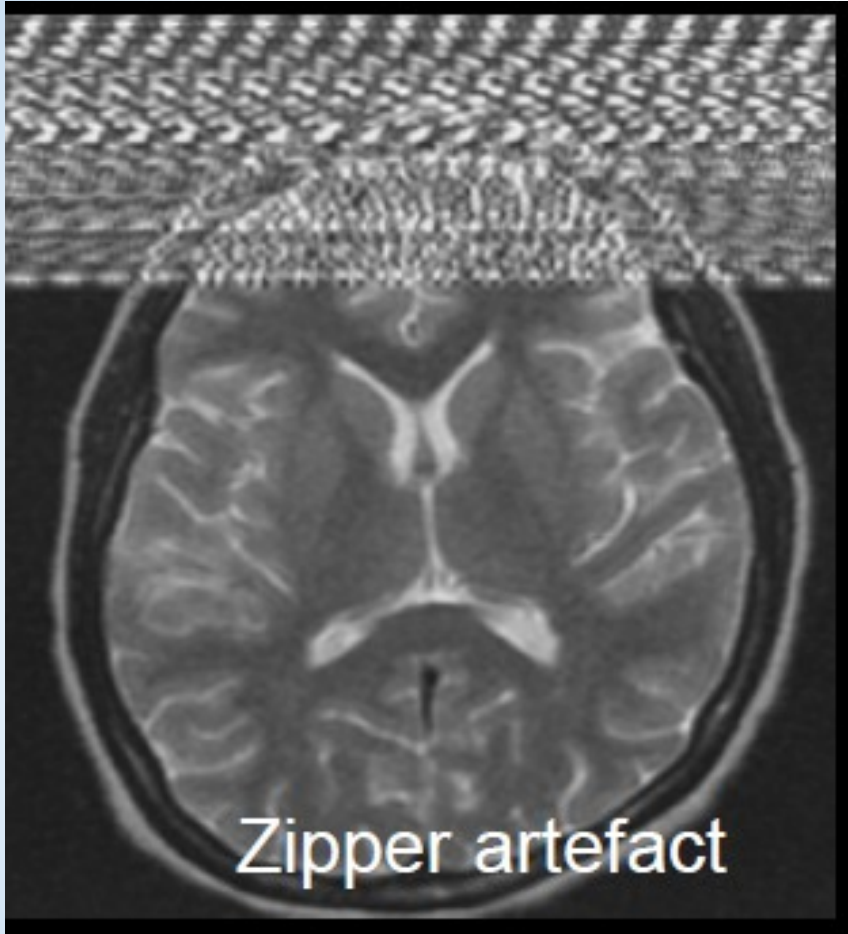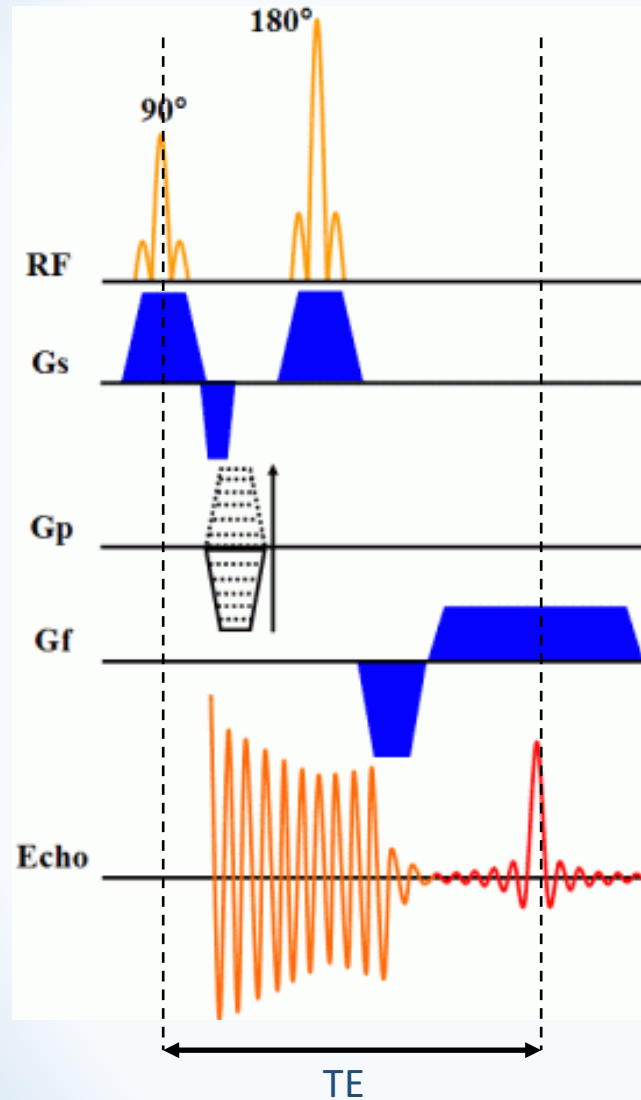
Sequences
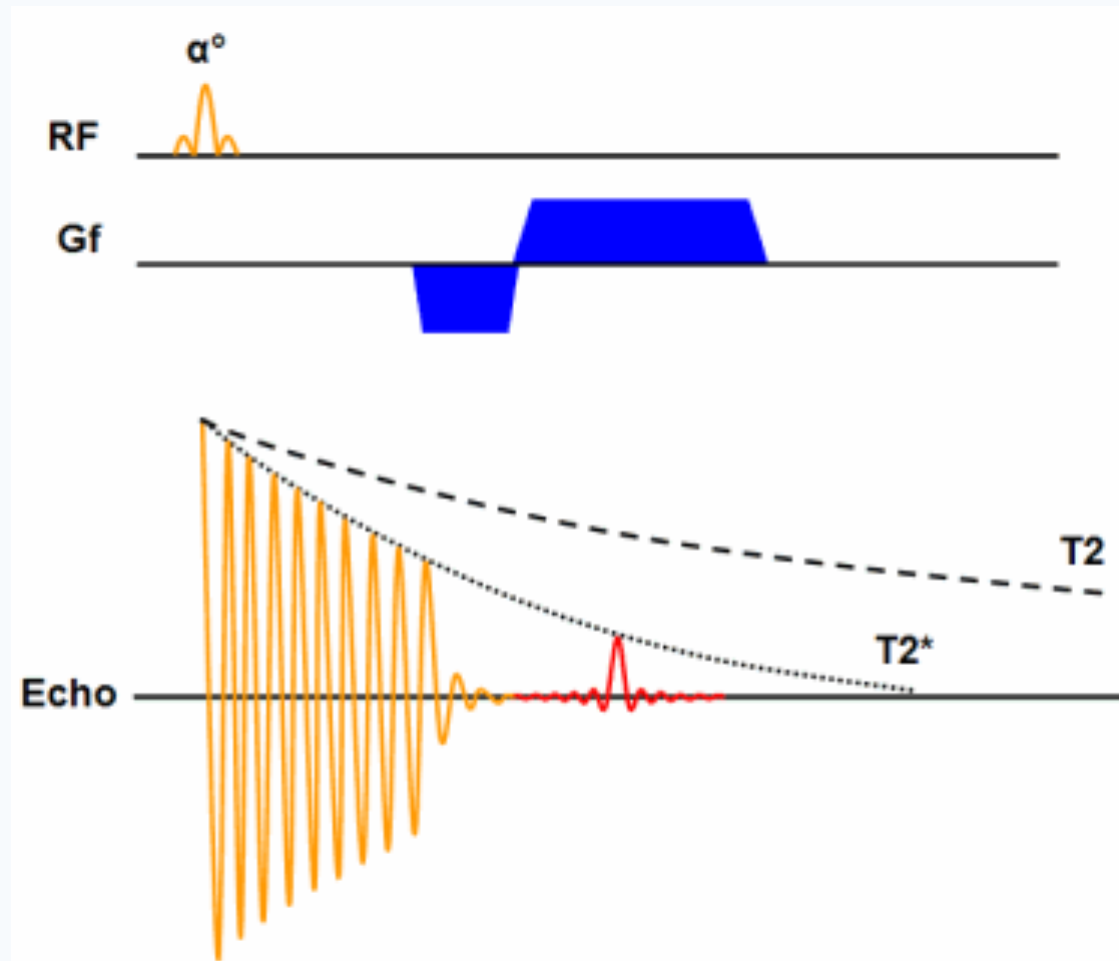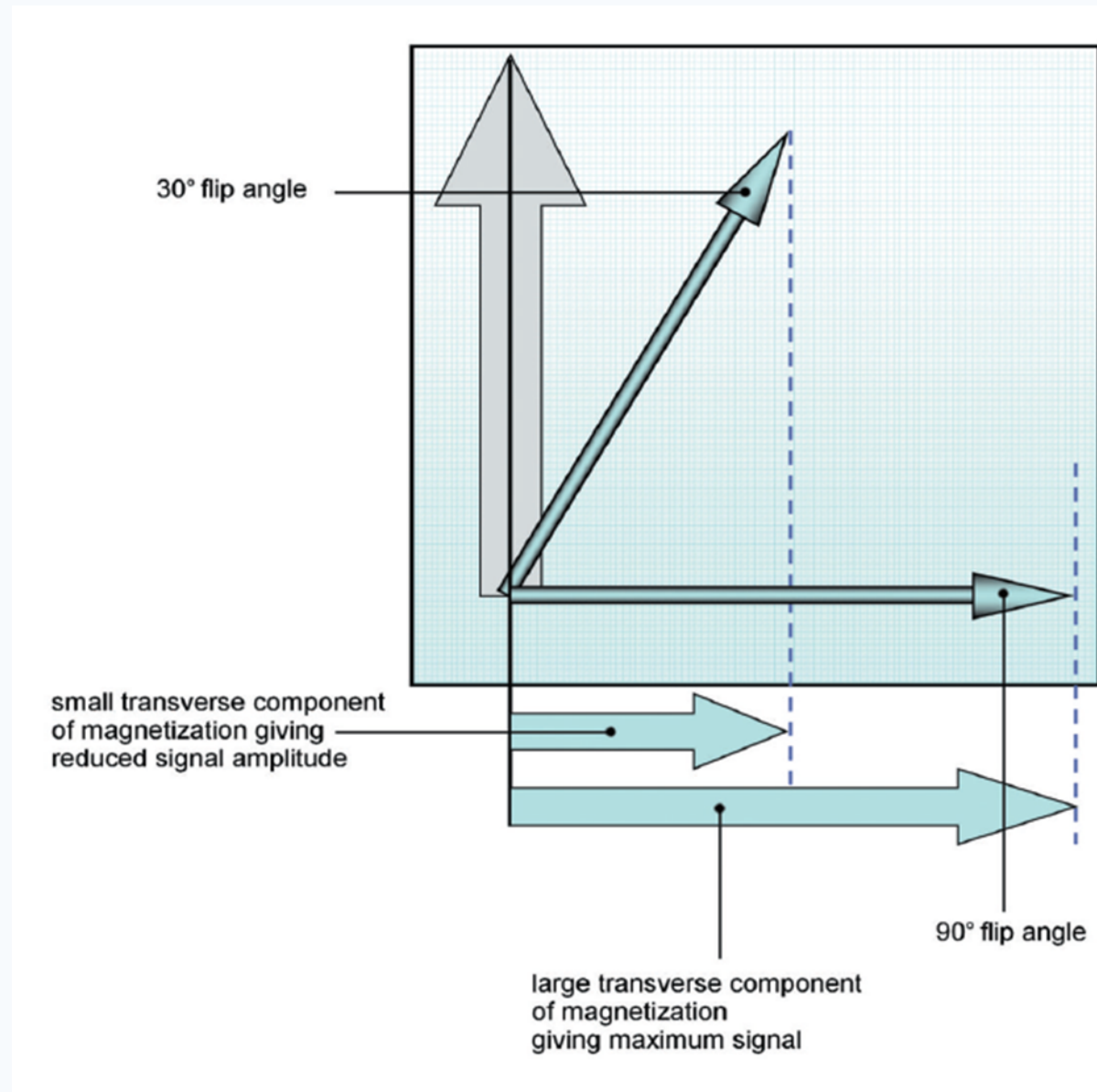
Signal from gradient echo is **much** weaker!

Very short TE, low Flip Angle (FA)
→ Very short TR is possible!

Finding optimal Flip Angle increases SNR of images

30° flip angle

small transverse component
of magnetization giving
reduced signal amplitude

90° flip angle

large transverse component
of magnetization
giving maximum signal

Spoiled gradient echo sequence FLASH (residual magnetization in transversal plane is destroyed using additional gradient)
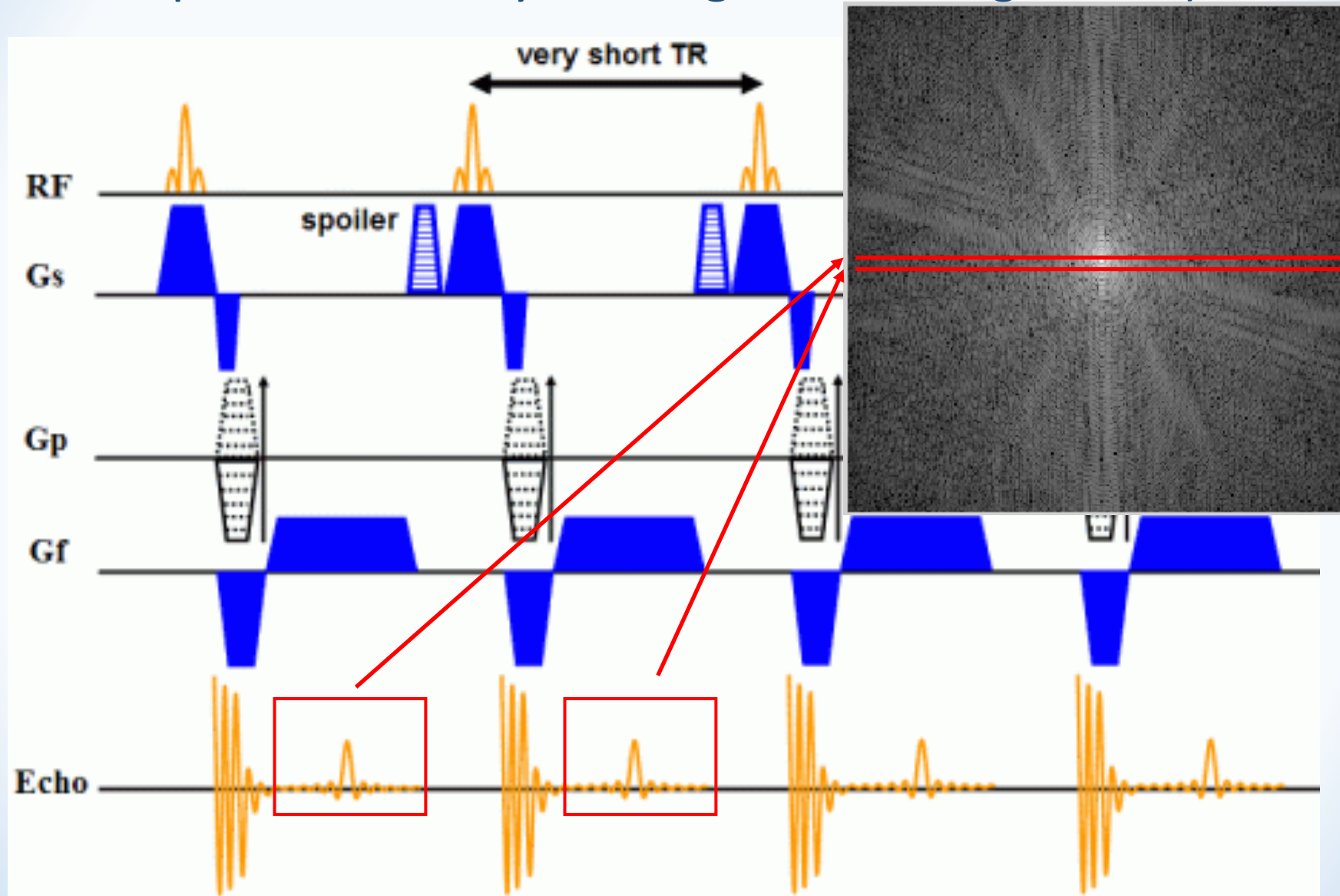
FLASH technique is a gradient echo sequence which combines a low-flip angle radio-frequency excitation of the NMR signal (recorded as a spatially encoded gradient echo) with a rapid repetition of the basic sequence. The repetition time is usually much shorter than the typical T1 relaxation time of the protons in biologic tissue. Only the combination of (i) a low-flip angle excitation which leaves unused longitudinal magnetization for an immediate next excitation with (ii) the acquisition of a gradient echo which does not need a further radio-frequency pulse that would affect the residual longitudinal magnetization, allows for the rapid repetition of the basic sequence interval and the resulting speed of the entire image acquisition.

FLASH sequence eliminates all waiting periods needed to accommodate effects from T1 saturation. FLASH reduces the typical sequence interval to what is minimally required for imaging: a slice-selective radio-frequency pulse and gradient, a phase-encoding gradient, and a (reversed) frequency-encoding gradient generating the echo for data acquisition.

TrueFISP (True fast imaging with steady state precession)
is a coherent technique that uses a fully balanced gradient waveform. The image
contrast with TrueFISP is determined by T2*/T1 properties and mostly depending
on TR. The speed and relative motion insensitivity of acquisition help to make
the technique reliable, even in patients who have difficulty with holding their
breath.
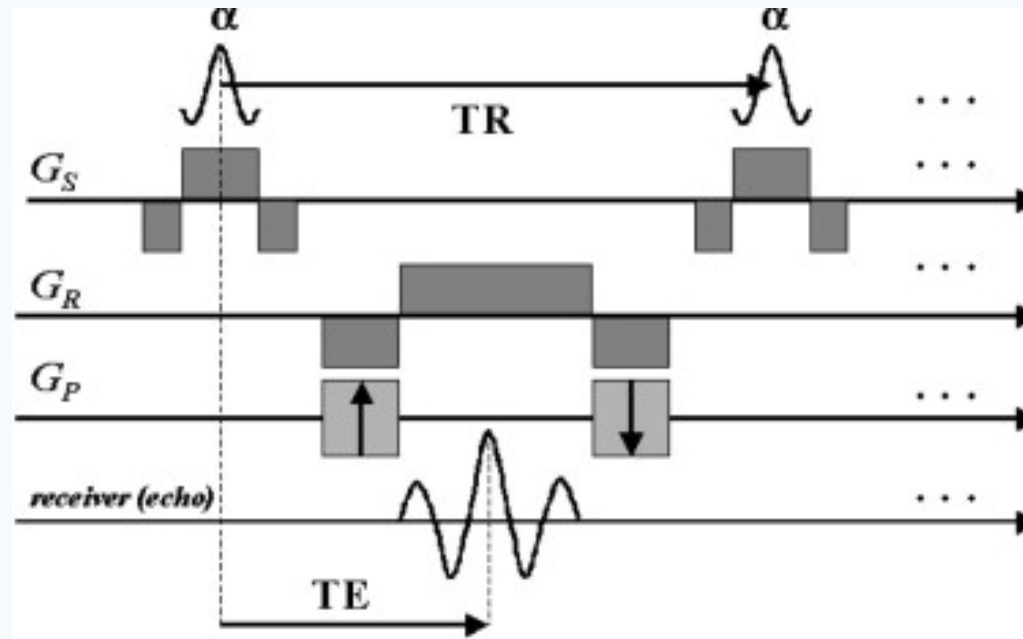
Usually very good SNR, but prone to artefacts and high SAR.

Spin Echo sequence
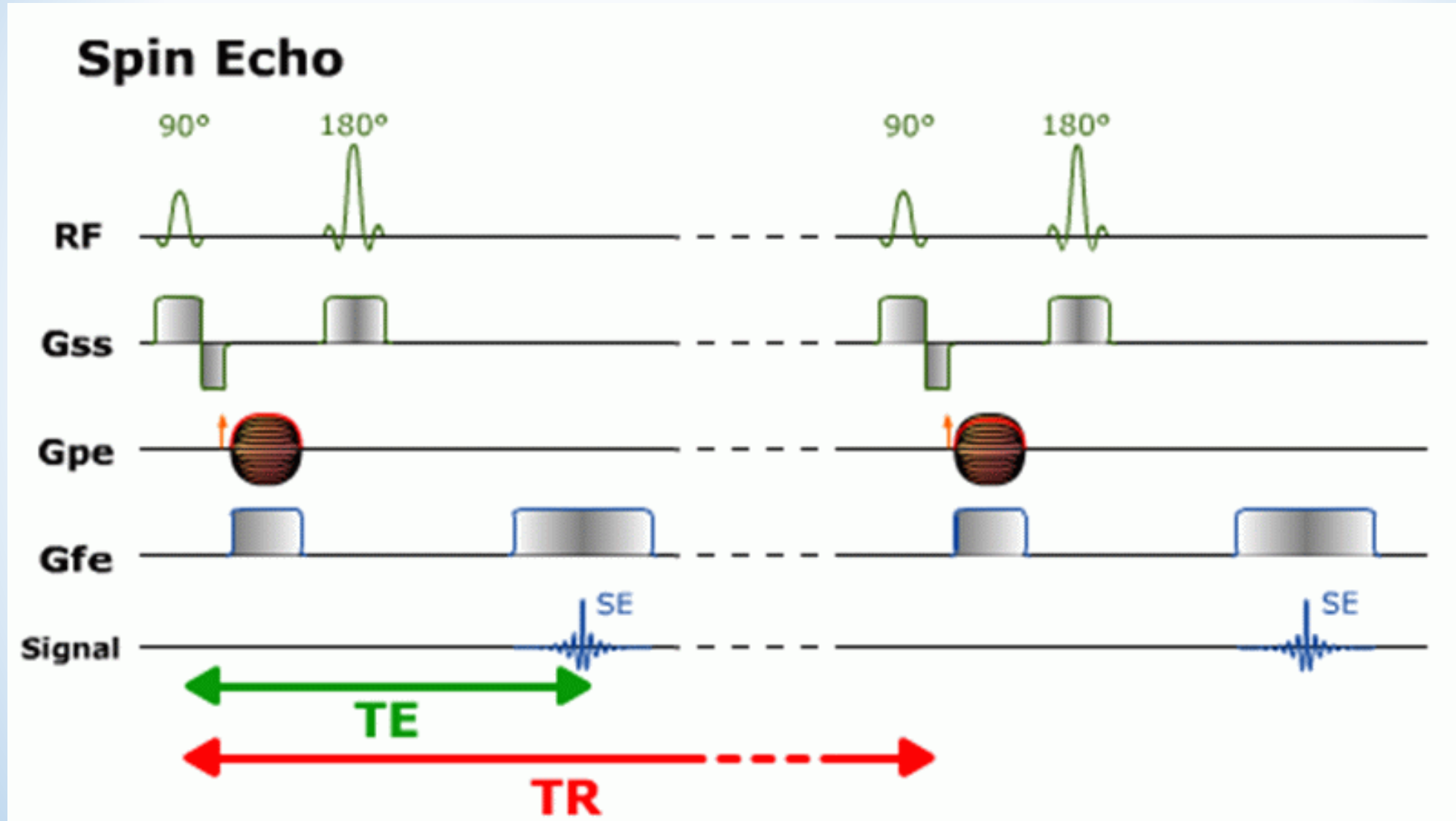
Fast Spin Echo "FSE" sequence (Turbo Spin Echo "TSE")

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.
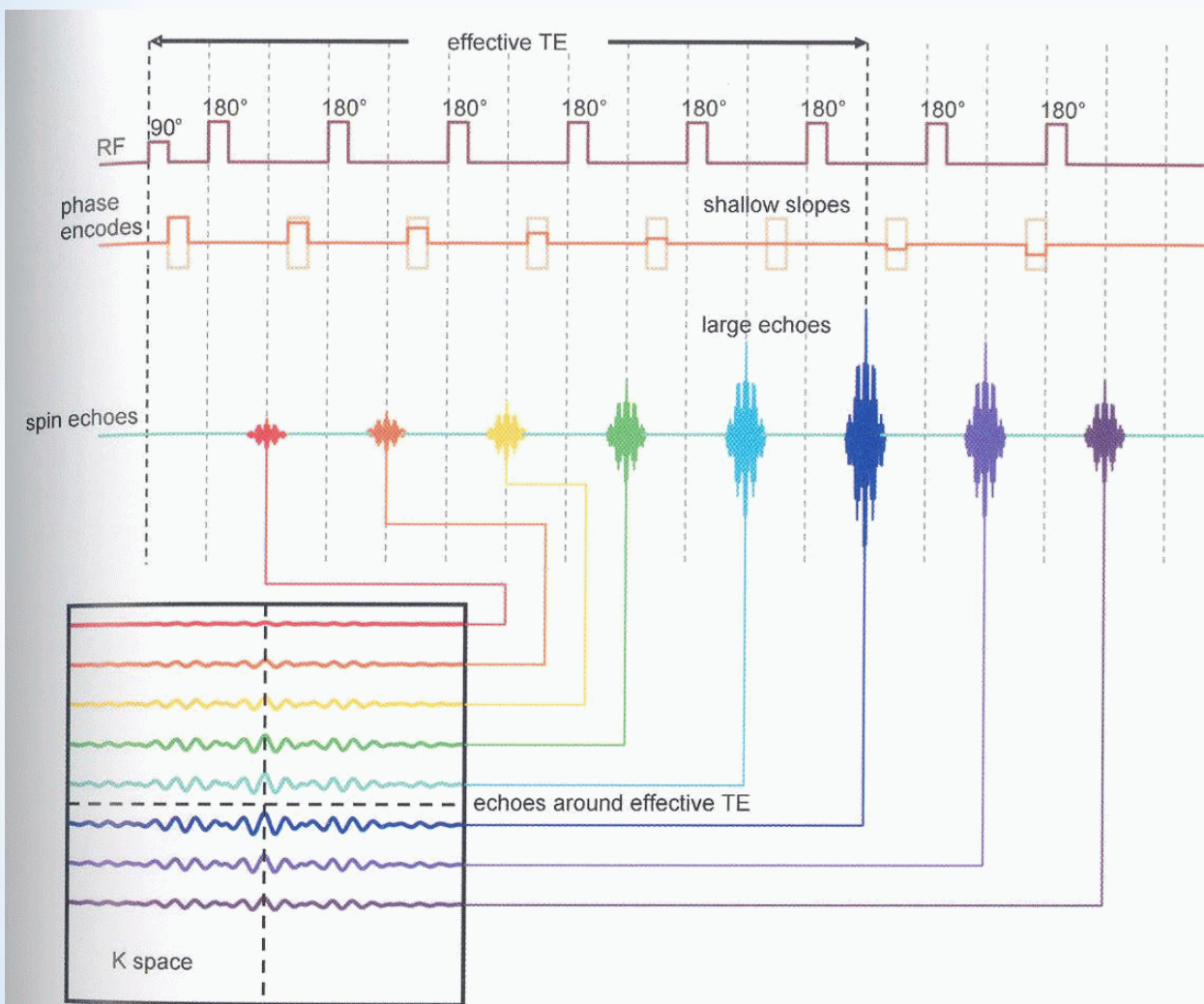
This can be repeated several times (more than 20 times in T2-weighted sequences).
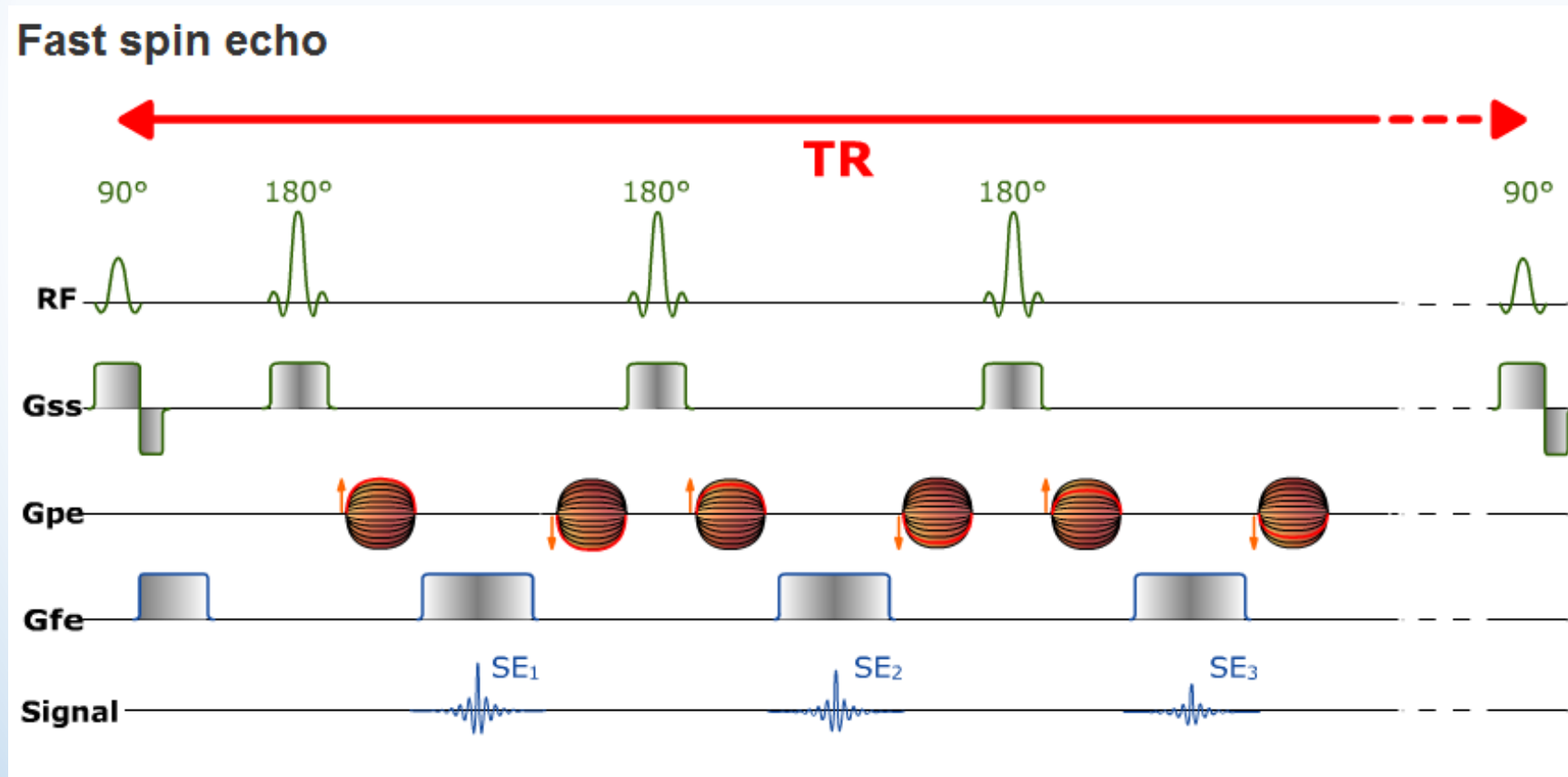
Within one TR several echoes are collected to k-space
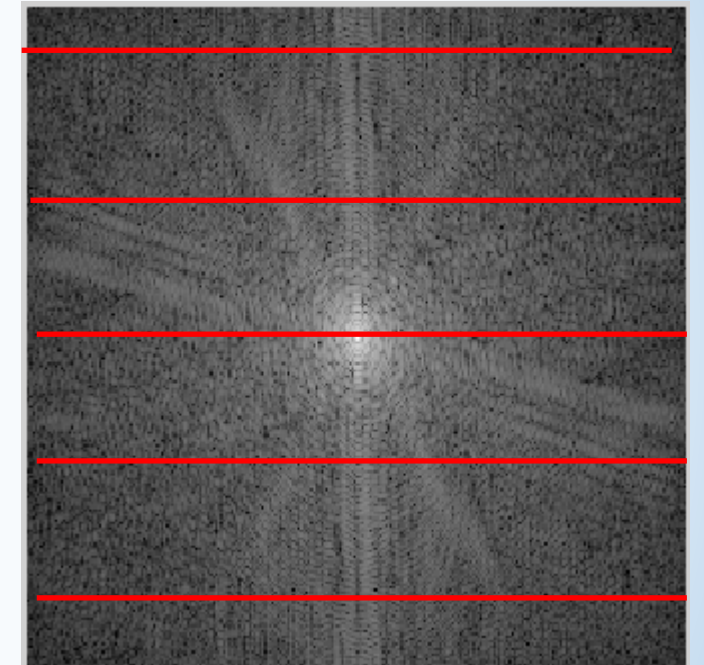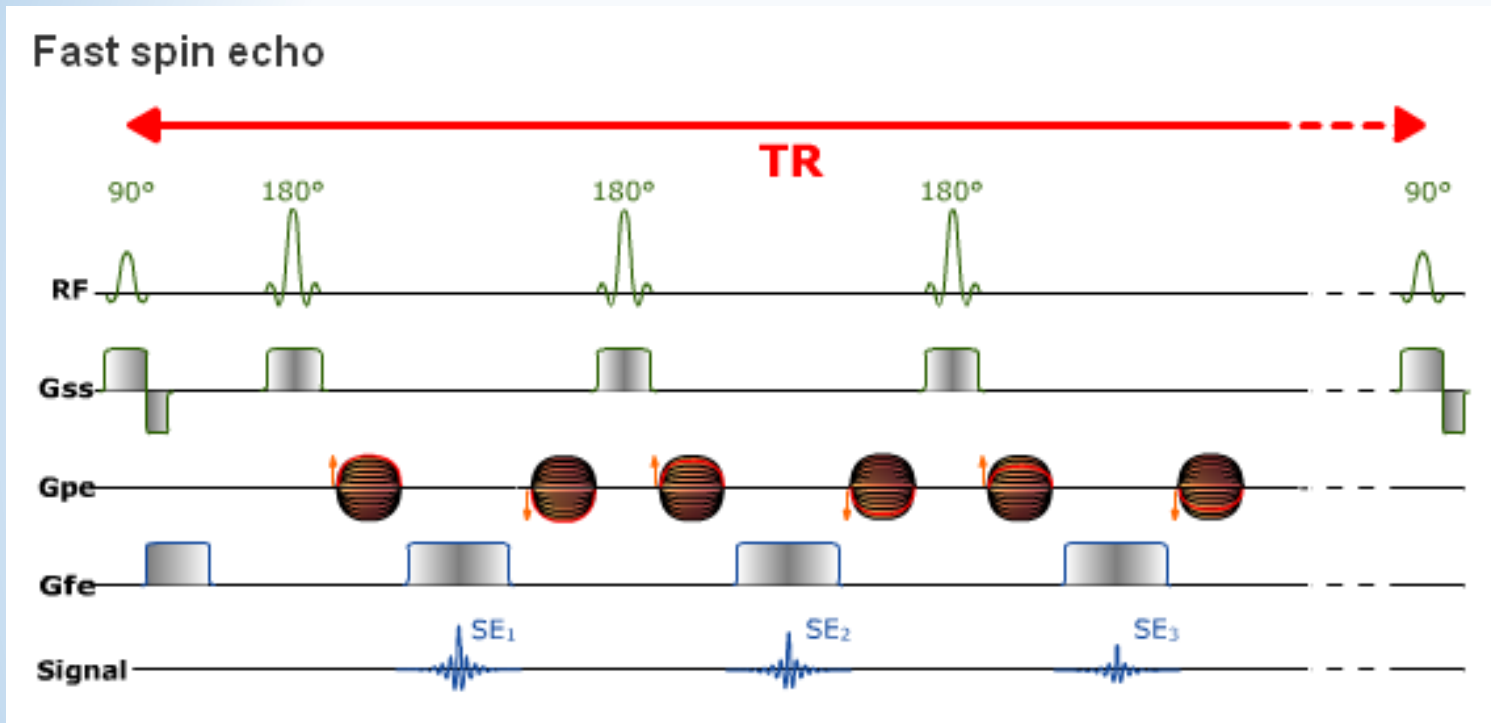
Contrast in FSE is based on "effective TE"

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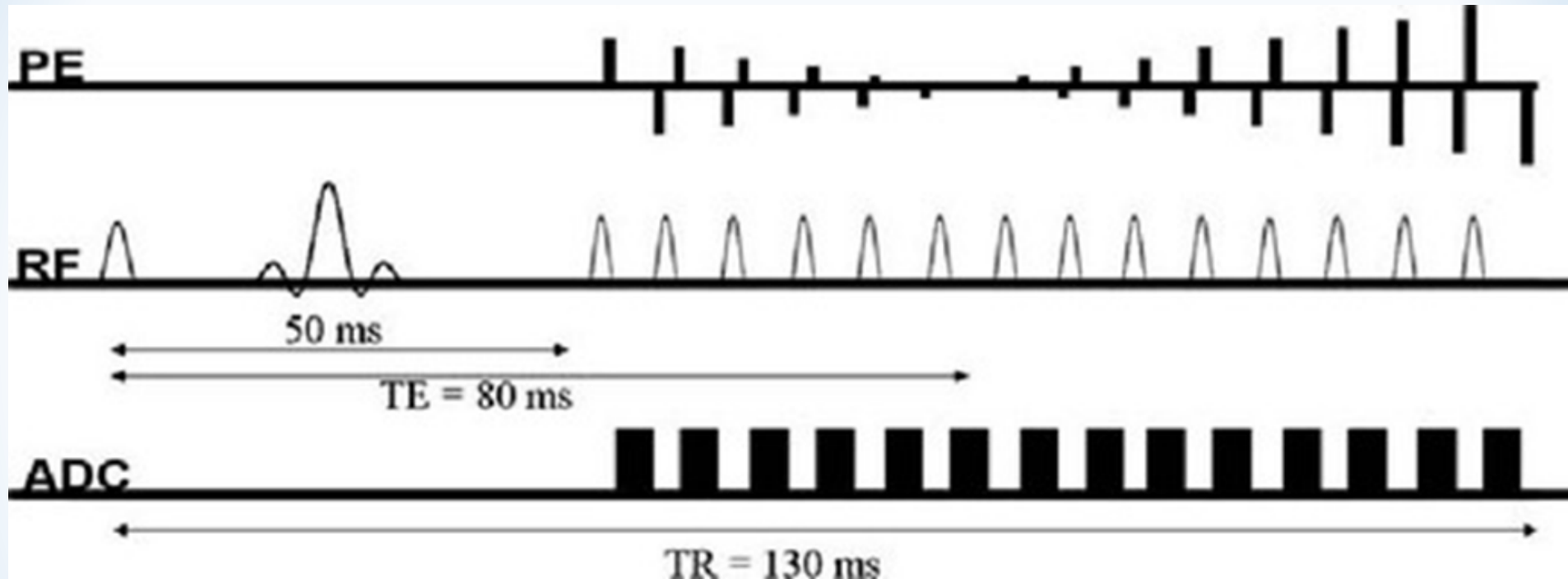
For example:
In PD sequence echoes with 10-30 ms TE are in the center of k-space

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.
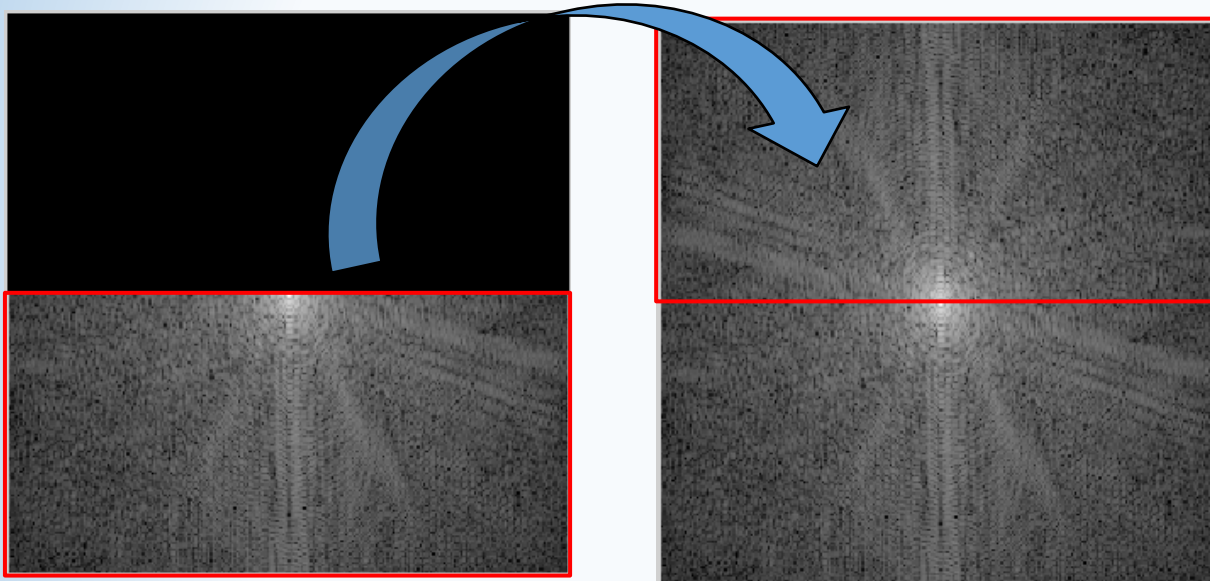
Provided no phase errors occur during data collection, k-space possesses a peculiar mirrored property known as conjugate (or Hermitian) symmetry. So, in theory, it is possible to create image using only half of the k-space data.
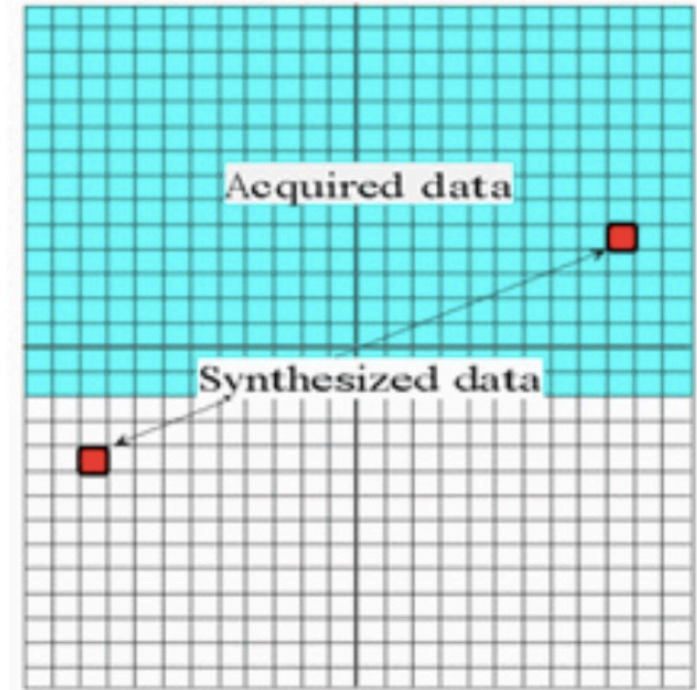
Phase-conjugate symmetry. About half of *k*-space is sampled by reducing the number of phase-encoding steps. The other half of *k*-space is synthesized/reconstructed.

Fast imaging techniques allow scan during breath-holds (10-20 s) or even in free-breathing
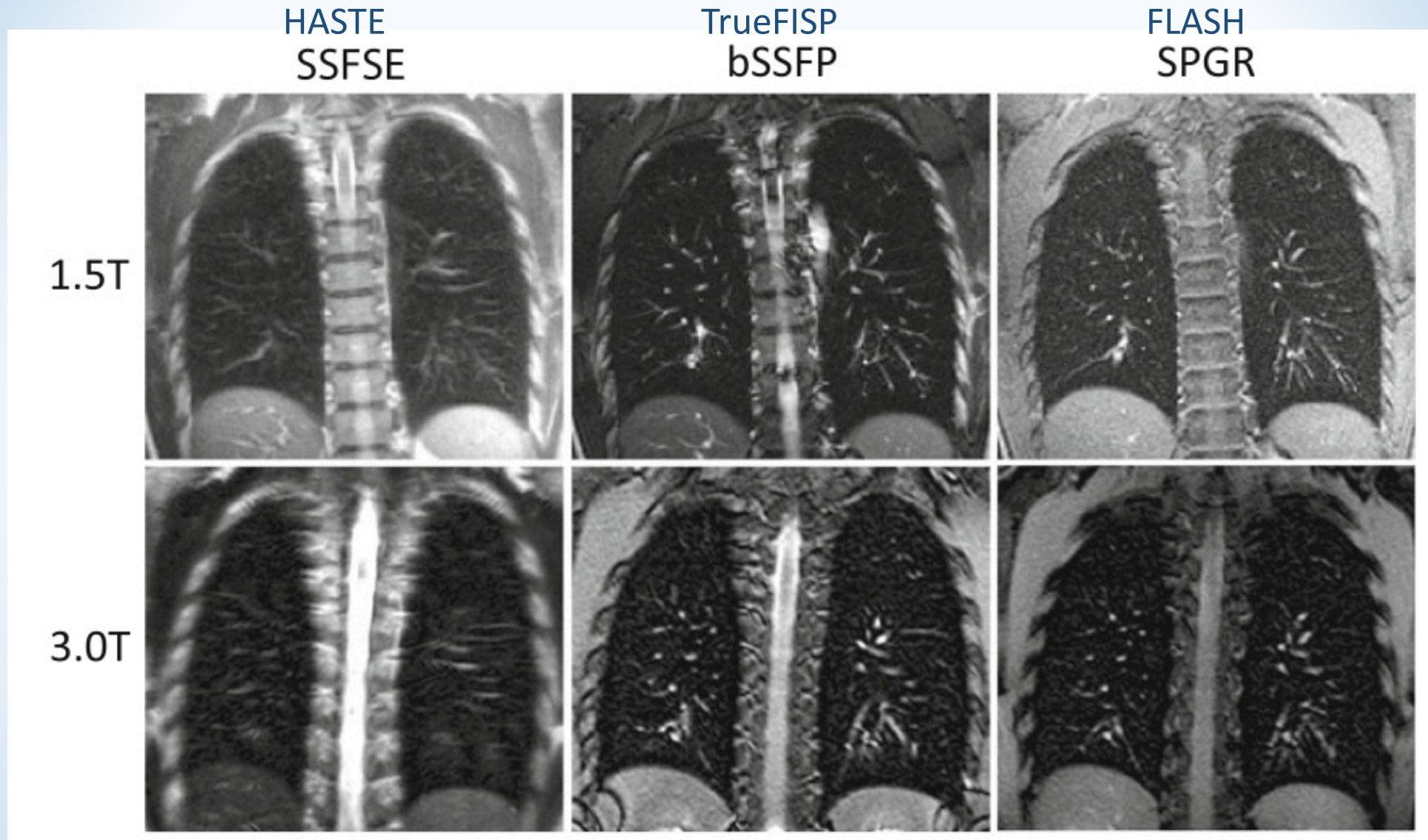
# Special techniques for motion reduction

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.
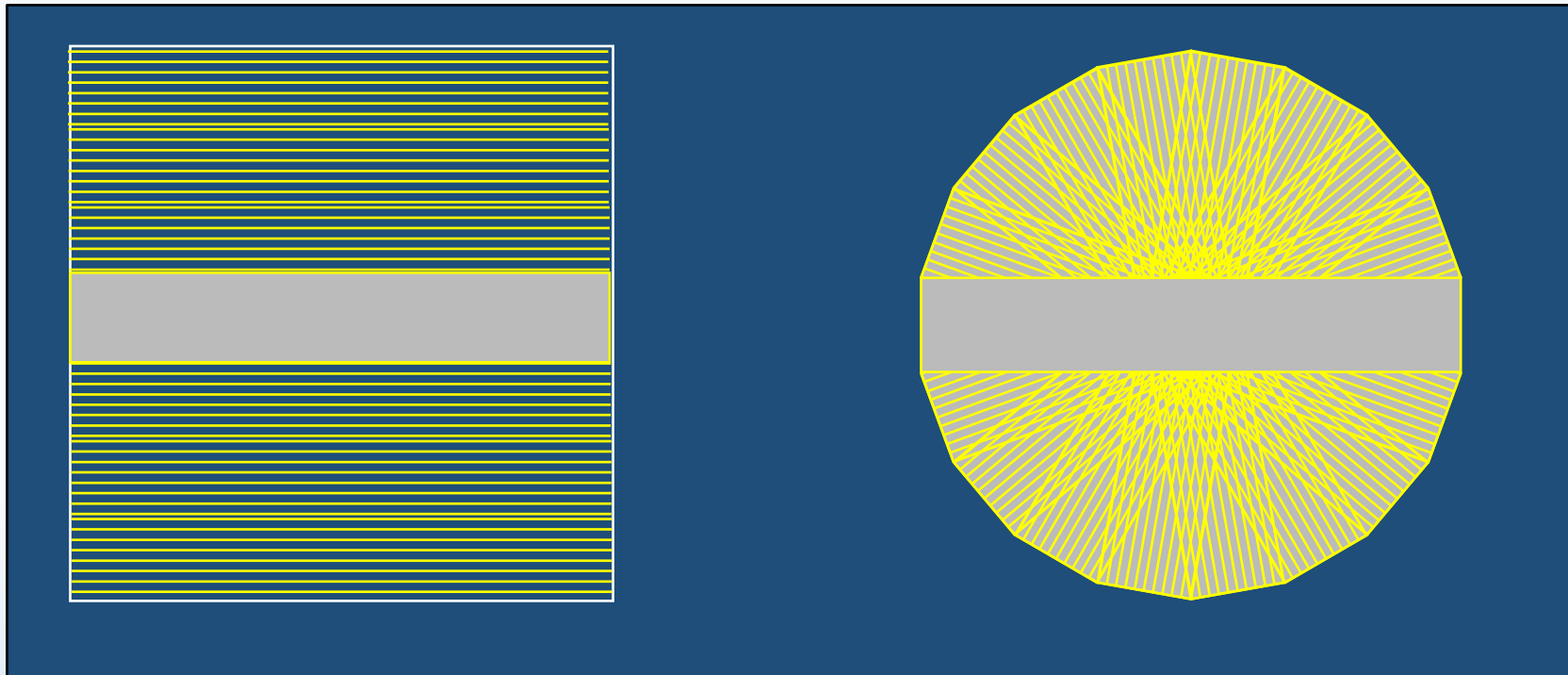
PROPELLER
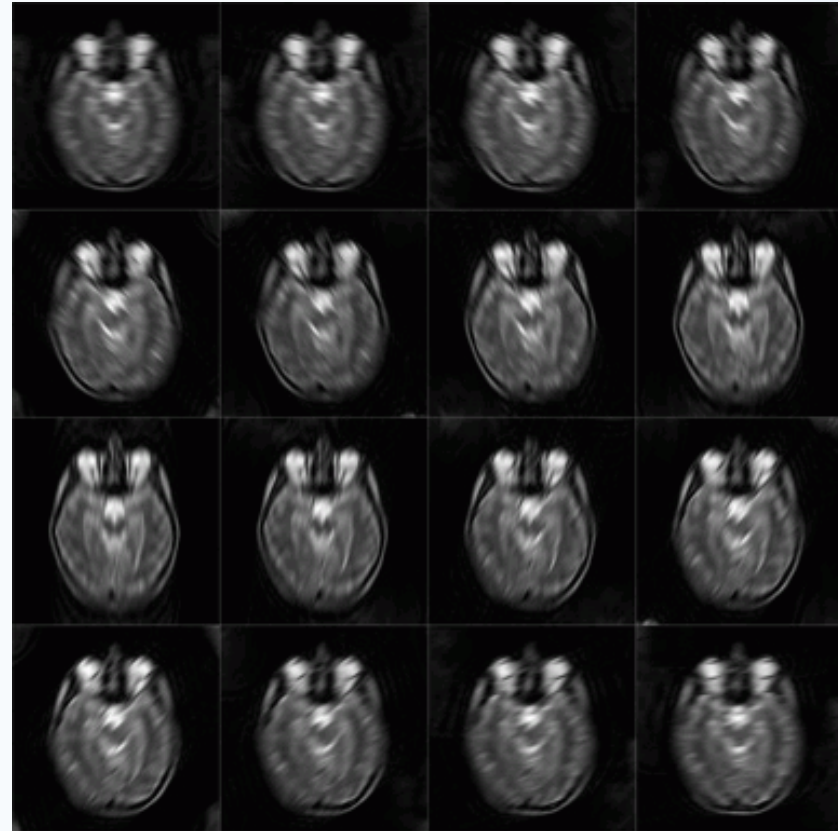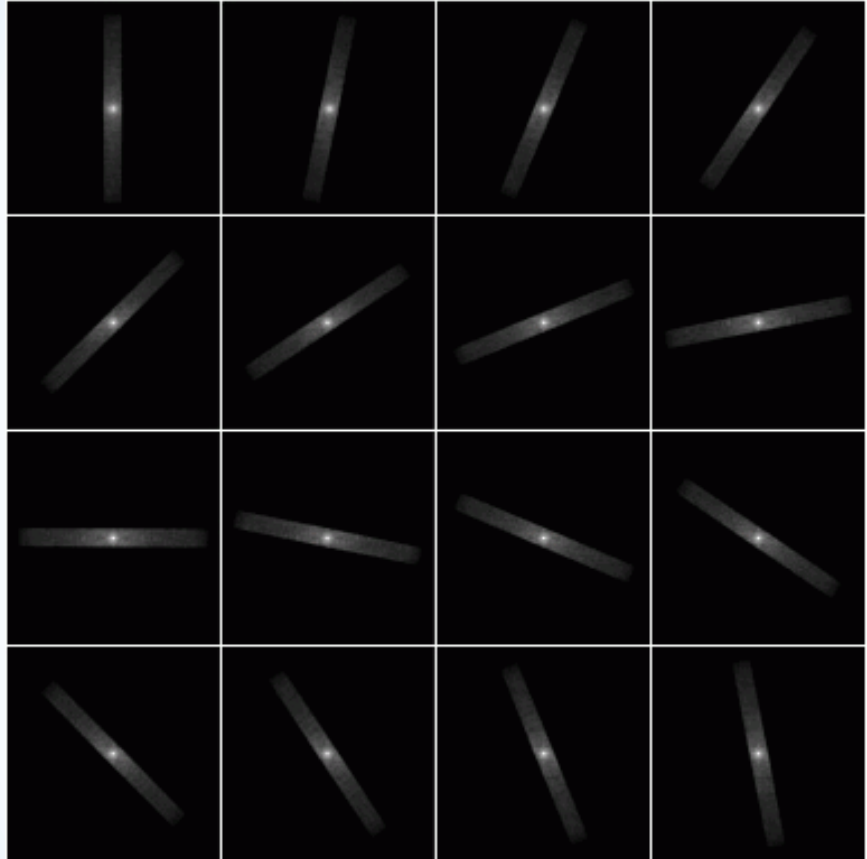
Regular data acquisition vs PROPELLER
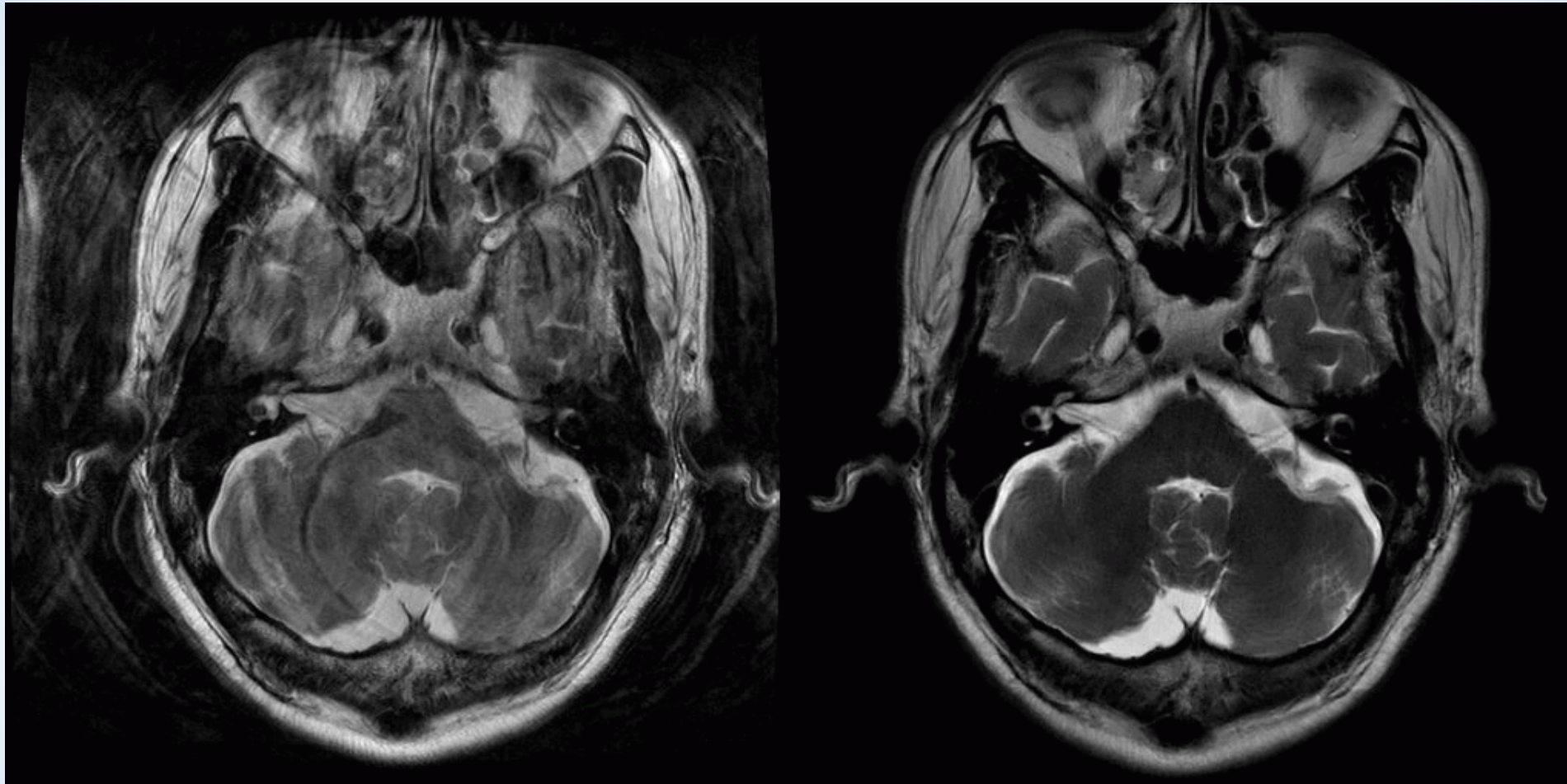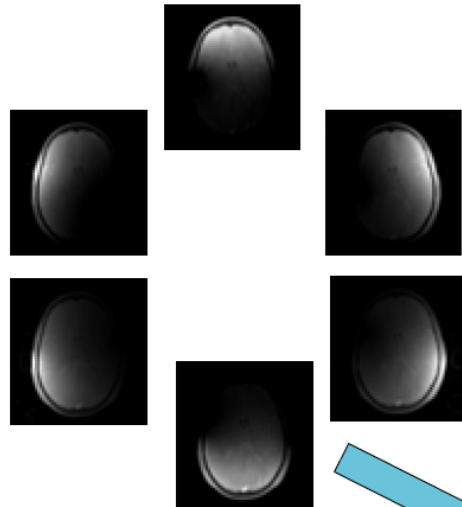
Parallel imaging

One more way to make k-space data acquisition faster.

Idea is to collect less lines (makes scan faster), but using this technique good resolution is maintained.
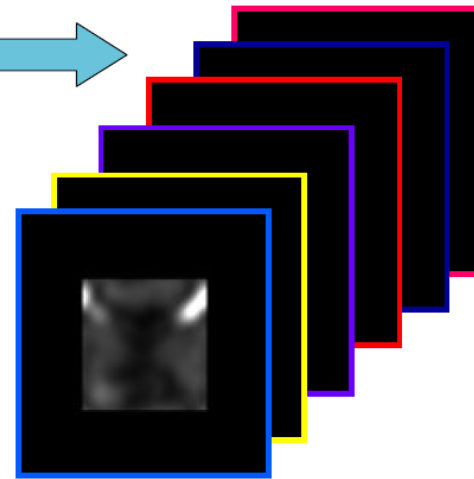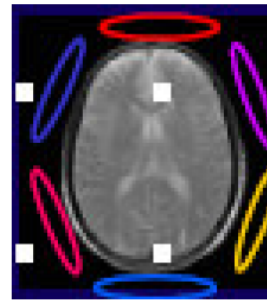
Problem is that there is still a loss of SNR.

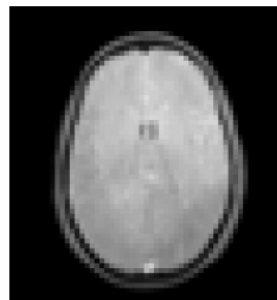# Receiver coils with several coil elements are demanded

SENSE technique

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
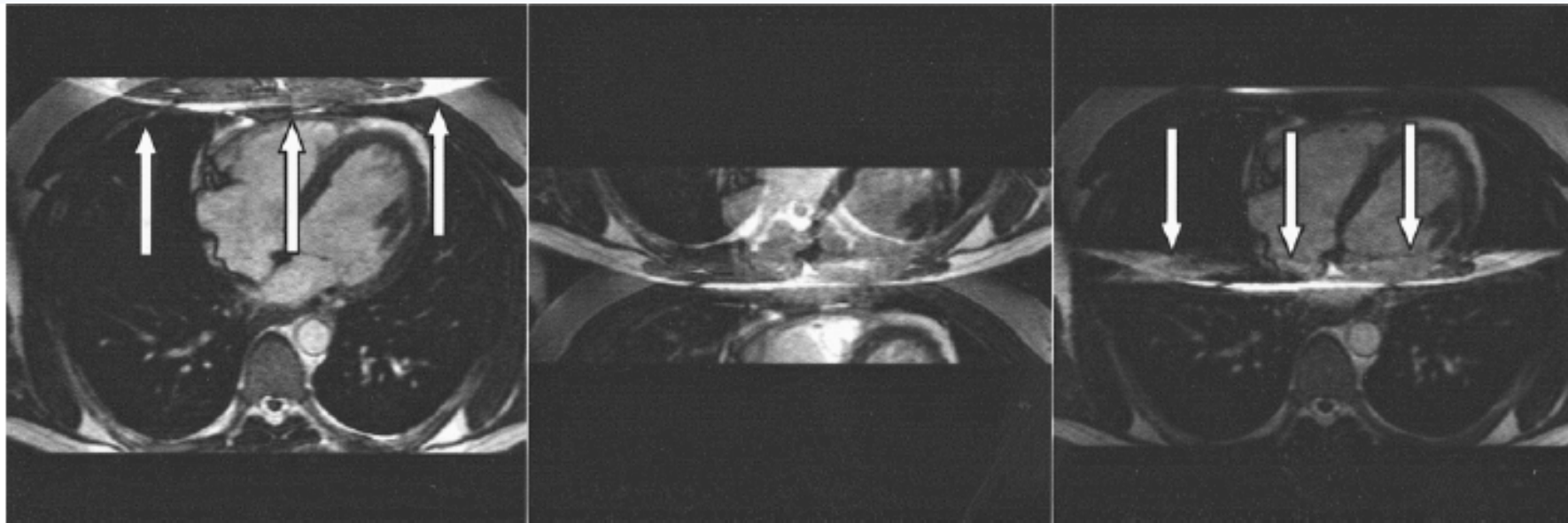
full FoV
without
aliasing

6.6.2018
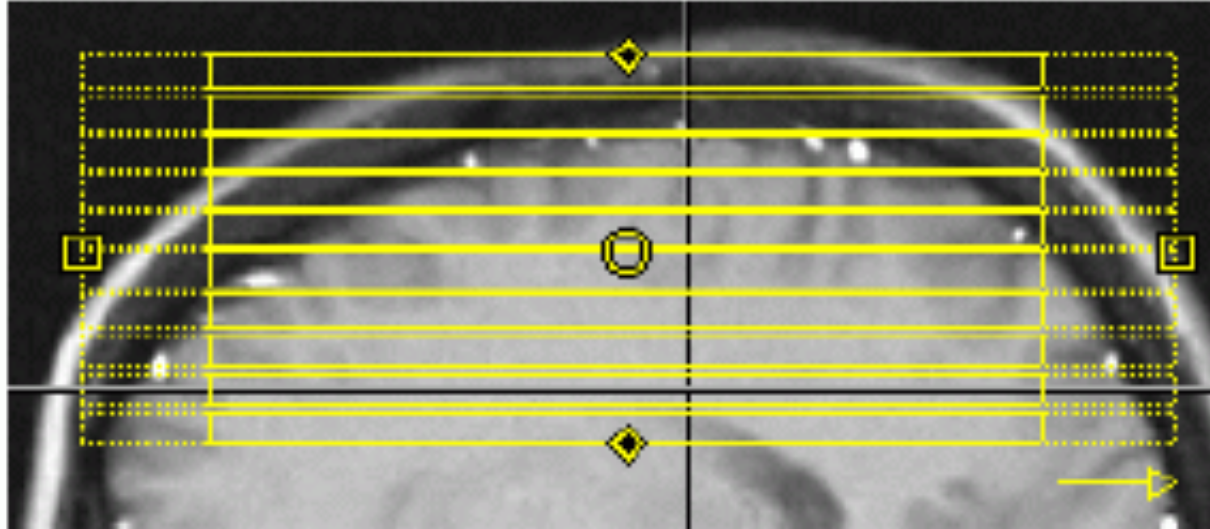
GRAPPA technique

6.6.2018
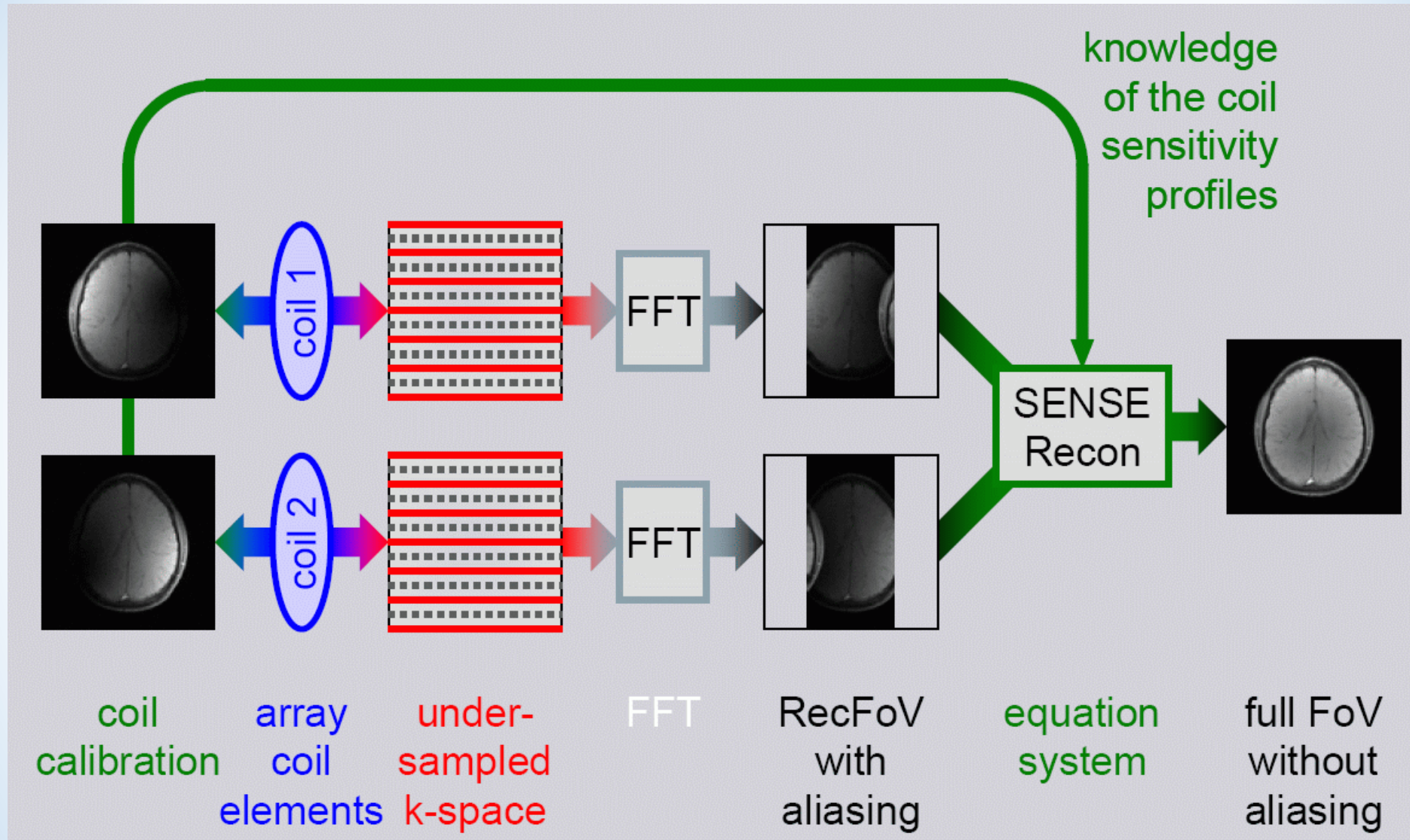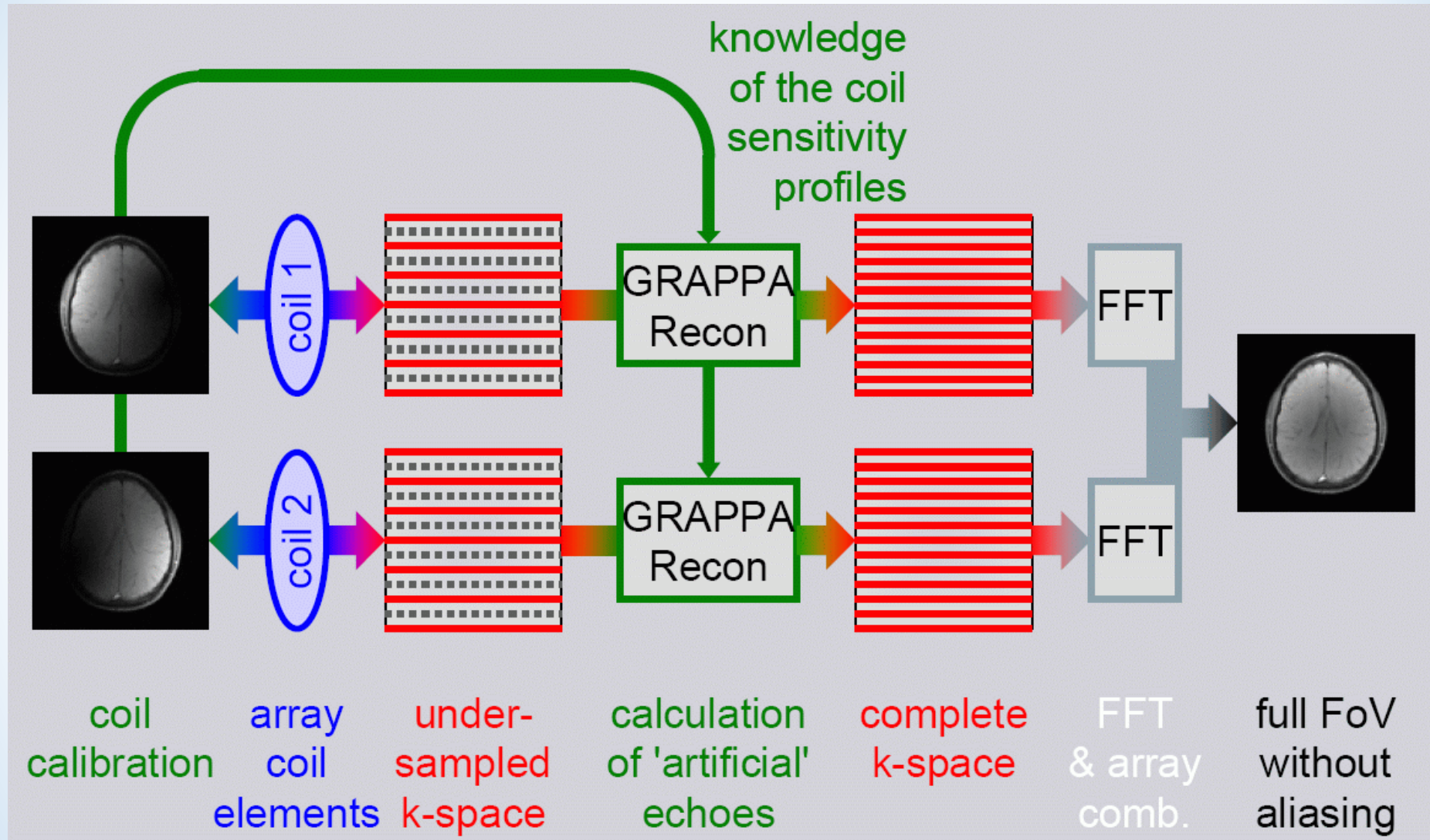
SENSE vs GRAPPA

GRAPPA technique is using additional information from reference lines already in k-space.

Wraparoun (aliasing) artefact is avoided when using GRAPPA.

Phase Oversampling / Foldover supression

This is how you avoid aliasing.

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

Phase Oversampling 0 -> 100%
SNR is increased by factor of 1.41

1/6/16