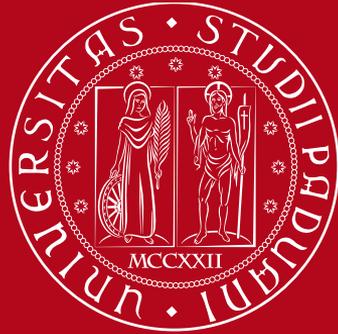


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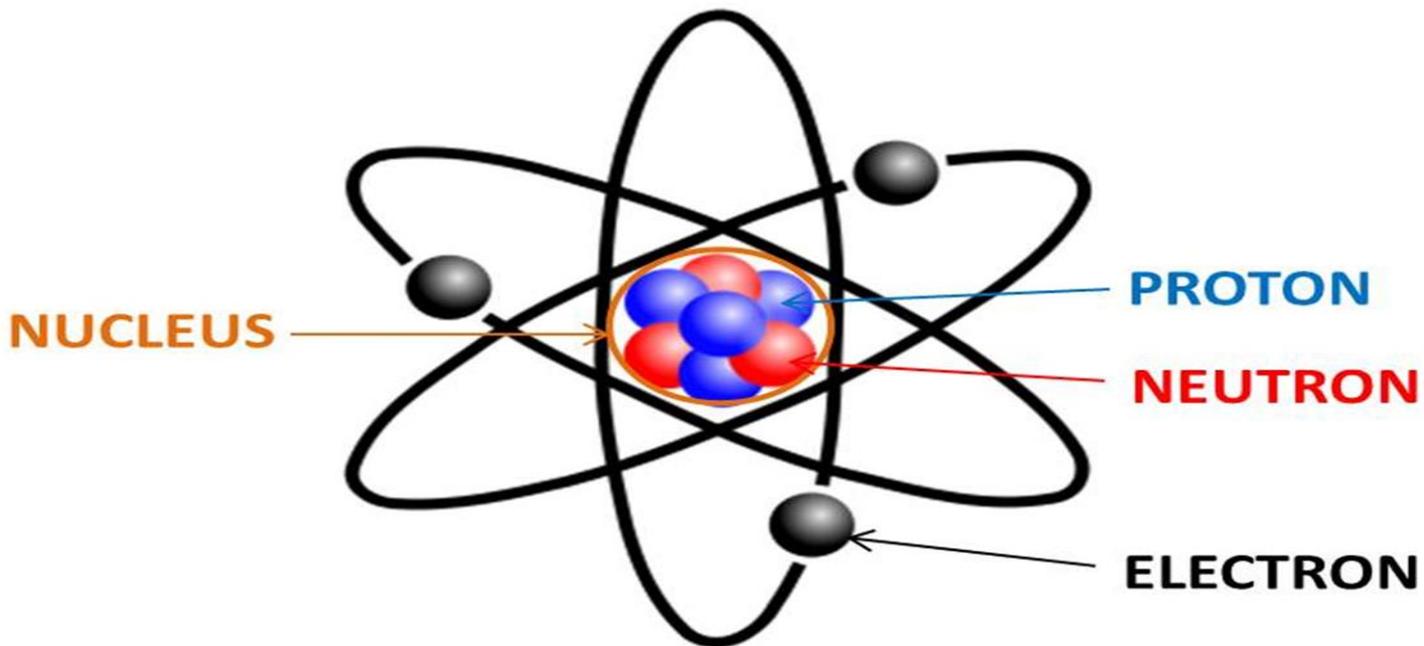


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PRINCIPI FISICI ED ACQUISIZIONE DEL SEGNALE IN RISONANZA MAGNETICA

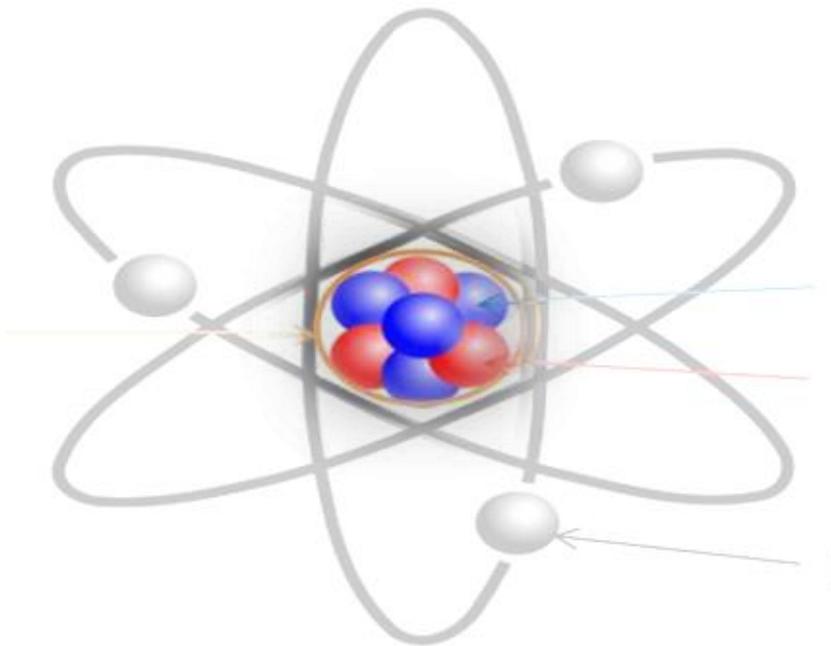
Manuel De Lazzari, MD PhD

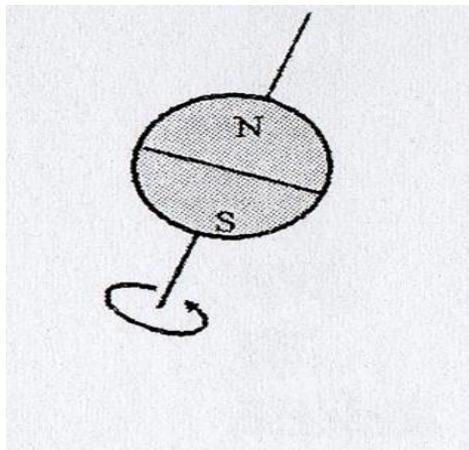
manuel.delazzari@aopd.veneto.it





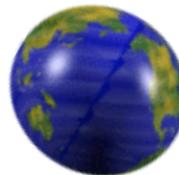
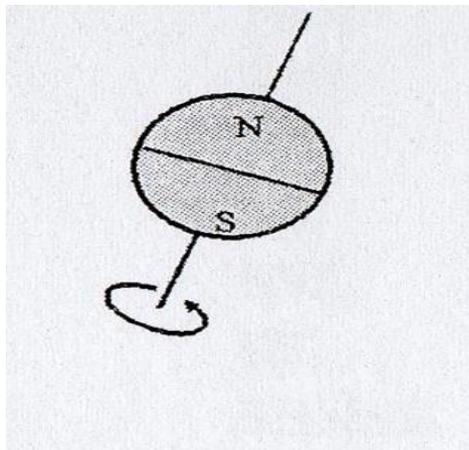
Atomic Nucleus





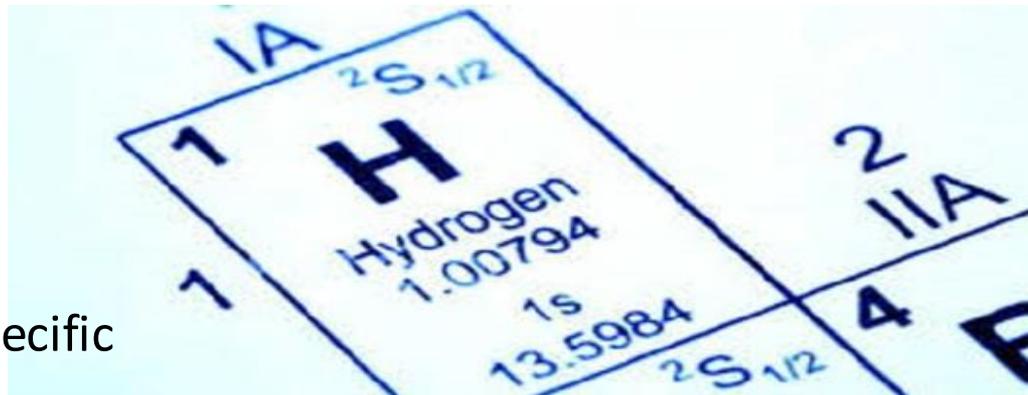
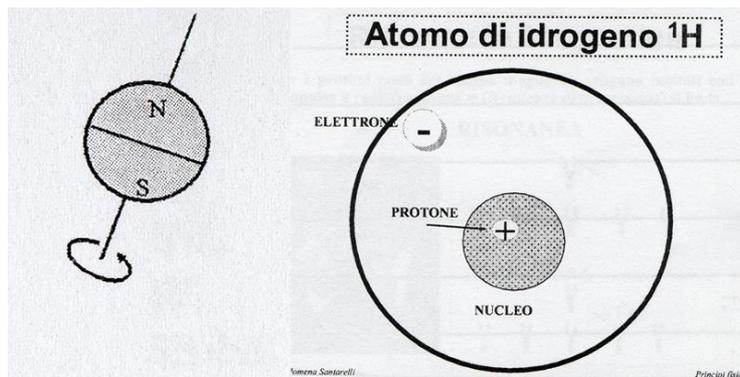
Nuclear spin is a property of atomic nuclei that depends on the numbers of neutrons and protons it contains.

Nuclei with an **unpaired neutrons or protons** have a non zero spin and so a **magnetic moment**.



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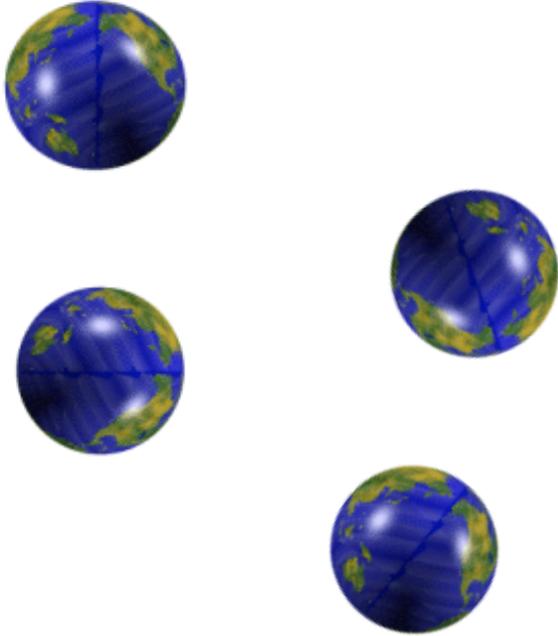


Gyromagnetic ratio (γ) is nucleus specific

i.e. for ^1H γ is 42.57 MHz/T



Nuclear Spin

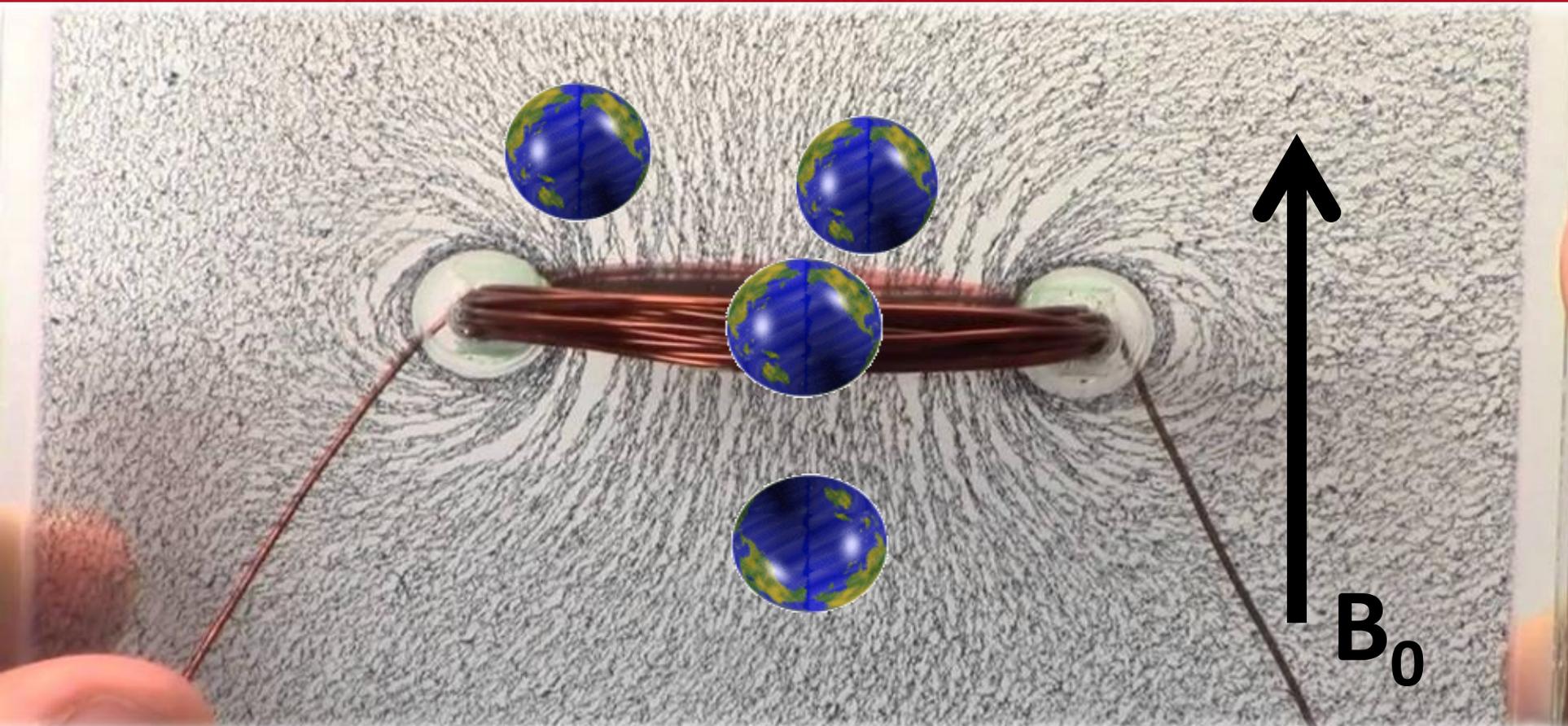


Normally, the direction of each magnetic moment vectors is randomly distributed.

The sum of all the spins gives a null net magnetization.

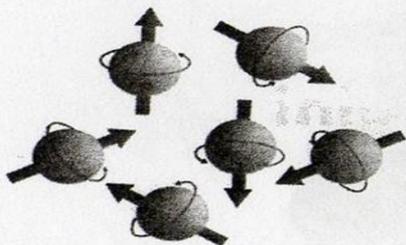


Nuclear Spin

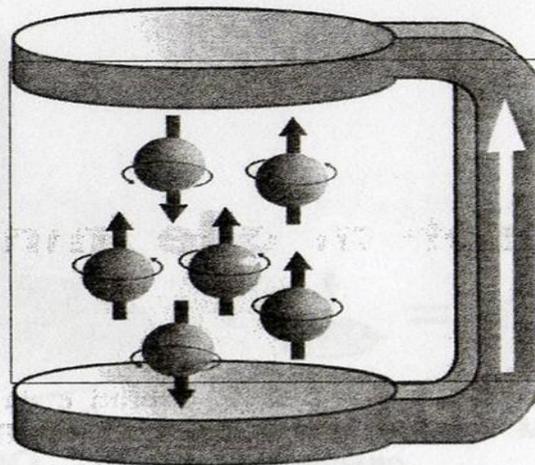




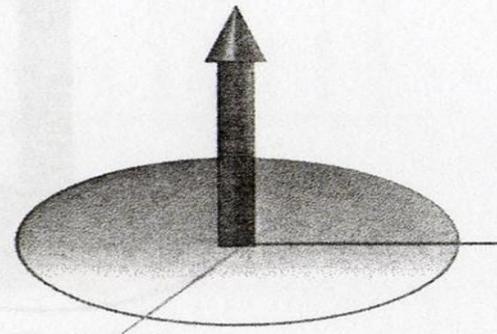
External Magnetic Field



Assi di rotazione orientati casualmente



Spin dei protoni allineati al campo magnetico



Vettore di magnetizzazione netta dei protoni in presenza di campo magnetico



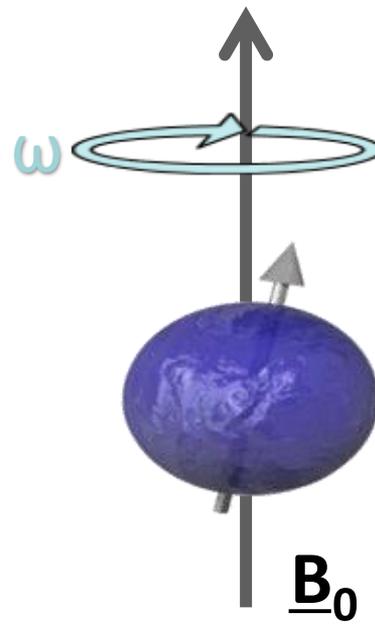
Larmor Frequency (ω)

$$\omega = \gamma B_0$$

ω = Larmor Frequency, Hertz

γ = gyromagnetic ratio (specific, for ^1H is 42.57 MHz/T)

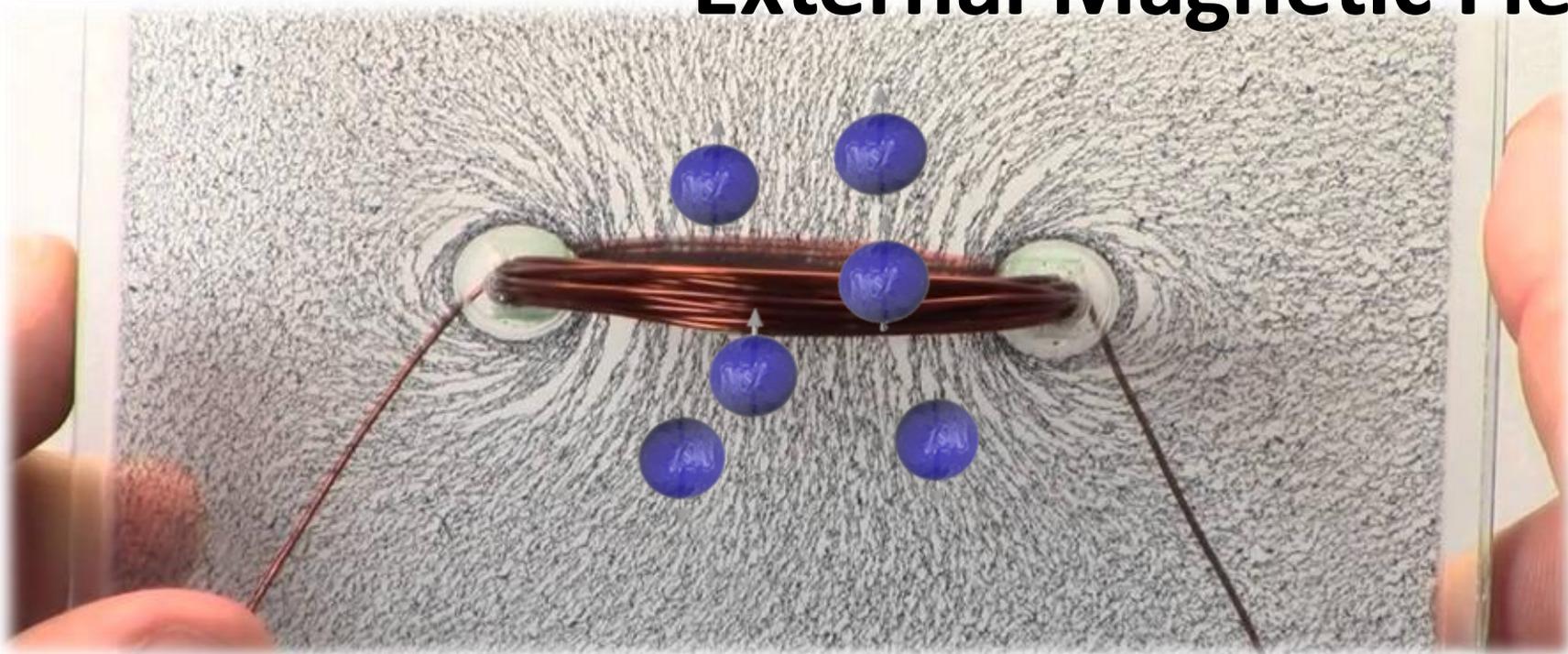
B_0 = magnetic field strength, Tesla





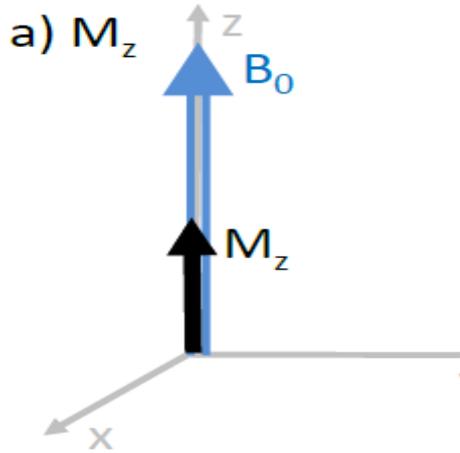
Nuclear Spin in

External Magnetic Field



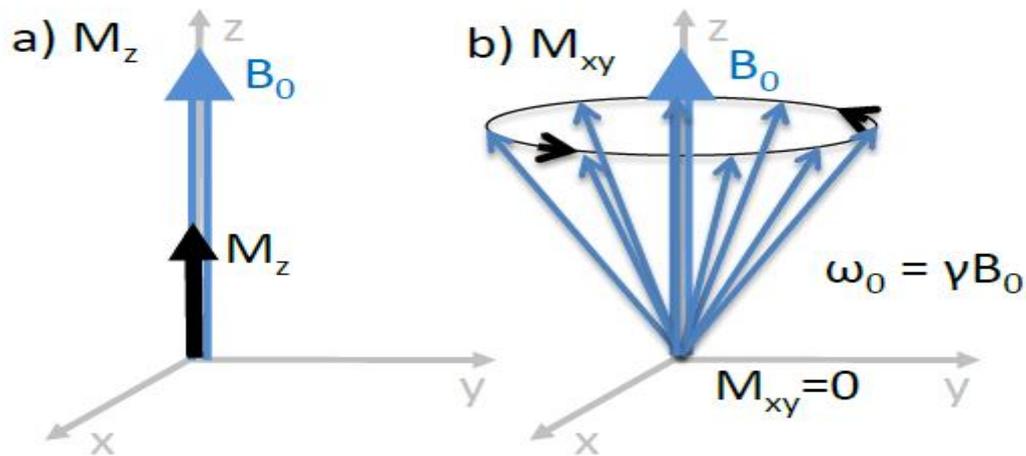


Precession and MR Signal



We measured not the amount of a single atom magnetisation vector but the sum of all (net magnetisation).

Precession and MR Signal



Into B_0 , the atoms exhibit a preference to align with the main magnetic field giving a net magnetisation M_z along z-axis

The precession depends upon B_0 but the **phase is random**, so the net transverse magnetisation along **XY** plane is zero.



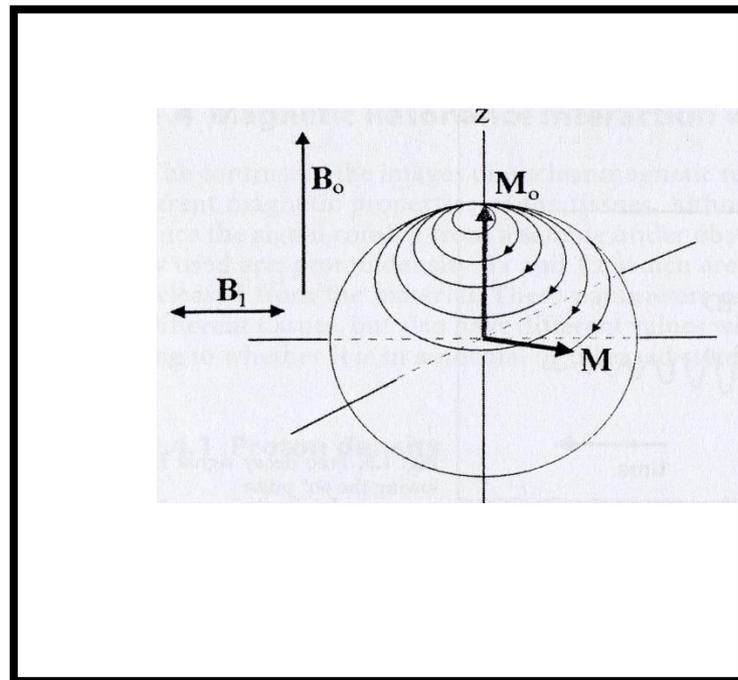
Disturbing the system

- **Resonance** corresponds to the energetic interaction between spins and electromagnetic radiofrequency (RF).



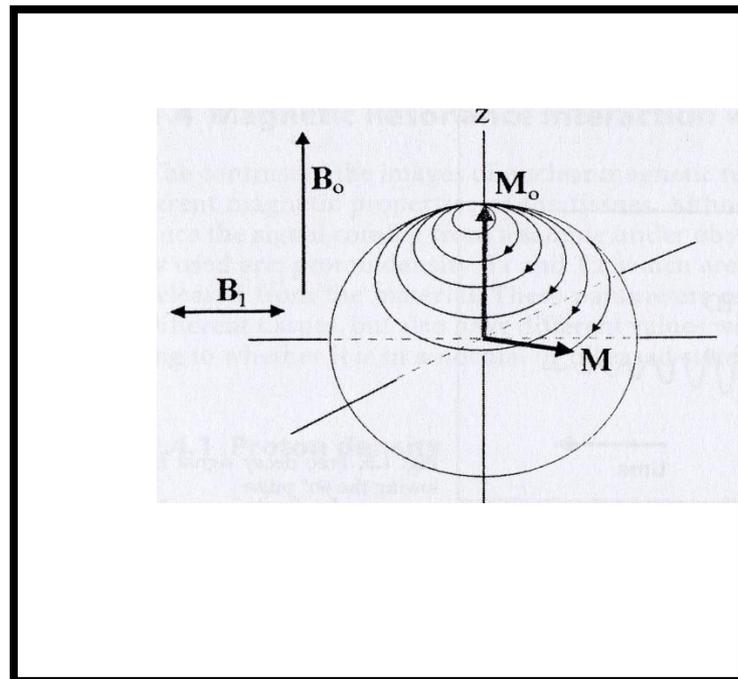
Radiofrequency

- Resonance corresponds to the energetic interaction between spins and electromagnetic radiofrequency (RF).
- An RF pulse is generated perpendicular to B_0 and revolving around B_0 with a frequency equal to ω .



Radiofrequency

- Resonance corresponds to the energetic interaction between spins and electromagnetic radiofrequency (RF).
- An RF pulse is generated perpendicular to B_0 and revolving around B_0 with a frequency equal to ω .
- RF field, B_1 , causes a **net magnetisation precession** about this field such that a component of the net magnetisation tips into the transverse plane.





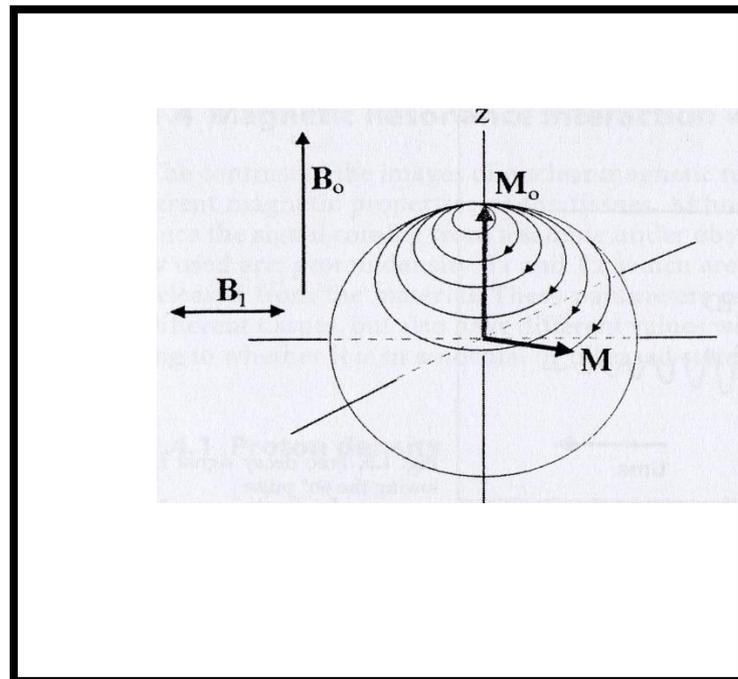
Radiofrequency

$$\alpha = \gamma B_1 t \quad \text{flip angle}$$

- t = time during B_1 works
- γ = gyromagnetic ratio

➤ After a 90° RF pulse, net magnetization tips down so that longitudinal magnetization has disappeared and transverse magnetization has appeared. **Transvers magnetization**

➤ After a 180° RF pulse, net magnetization became negative along Z axis. **Longitudinal magnetization**

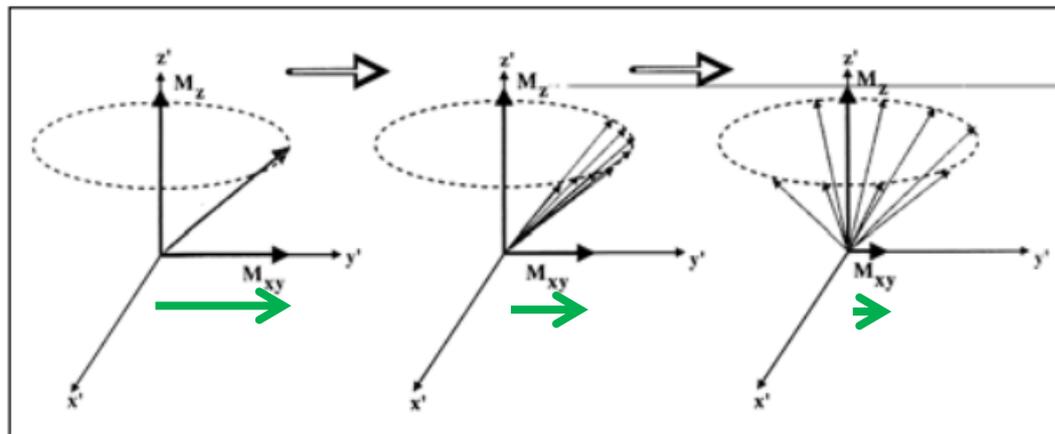




Free Induction Decay (FID)

After RF is turned off :

The net magnetization vector continued to precess about B_0 until to return at basal condition



Free Induction Decay (FID)

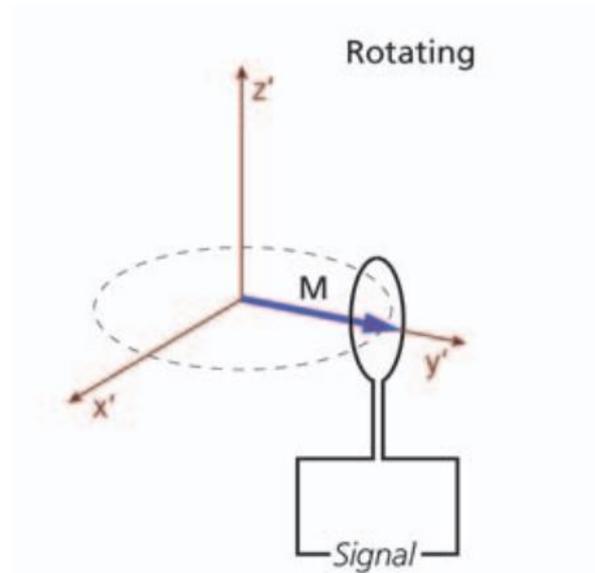
After RF is turned off :

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This moving magnetic vector induces a electric current in a **receive coil**.

This signal is recorded and represents the MR signal (FID)

Its amplitude is higher at begging, after decrease quickly because of field inhomogeneity.



Free Induction Decay (FID)

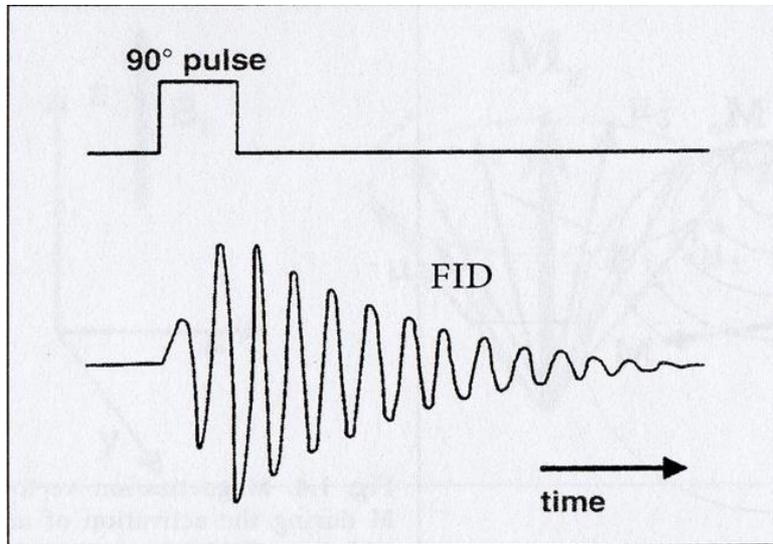
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- Basic Physics
- The creation of MR signal
- The acquisition of the signal



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- **What's?**

Is obtained by magnetic field gradients added to B_0 but applied at distinct moments and in different directions

- **Why?**

To localize the voxel signal

- 1. Slice selection gradient**
- 2. Phase encoding gradient**
- 3. Frequency encoding gradient**

Image Localization

- **Slice selection (z-axis):** a magnetic field gradient, Slice Selection Gradient (G_{SS}), is applied added to B_0 . The protons present a resonance frequency variation proportionate to G_{SS} according to Larmor equation.
- Simultaneously a RF is applied with the same frequency of the protons in the desired slice plan

$$\omega = \gamma B_0$$

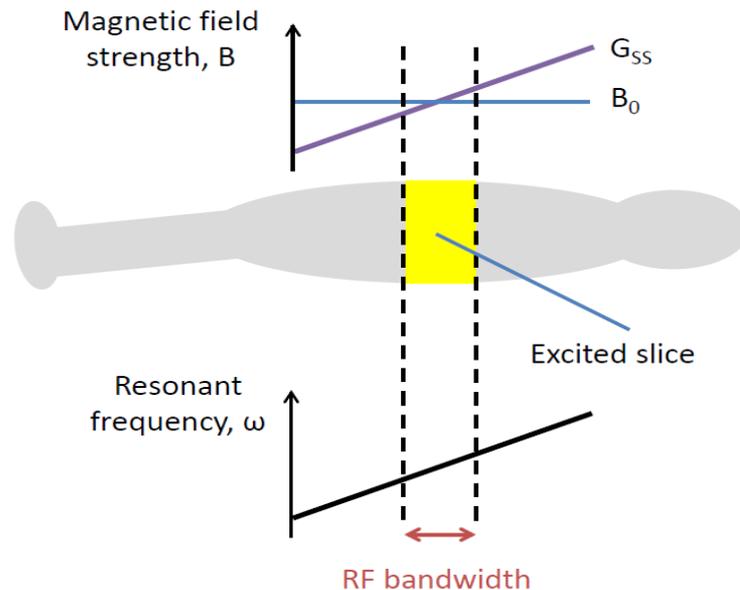


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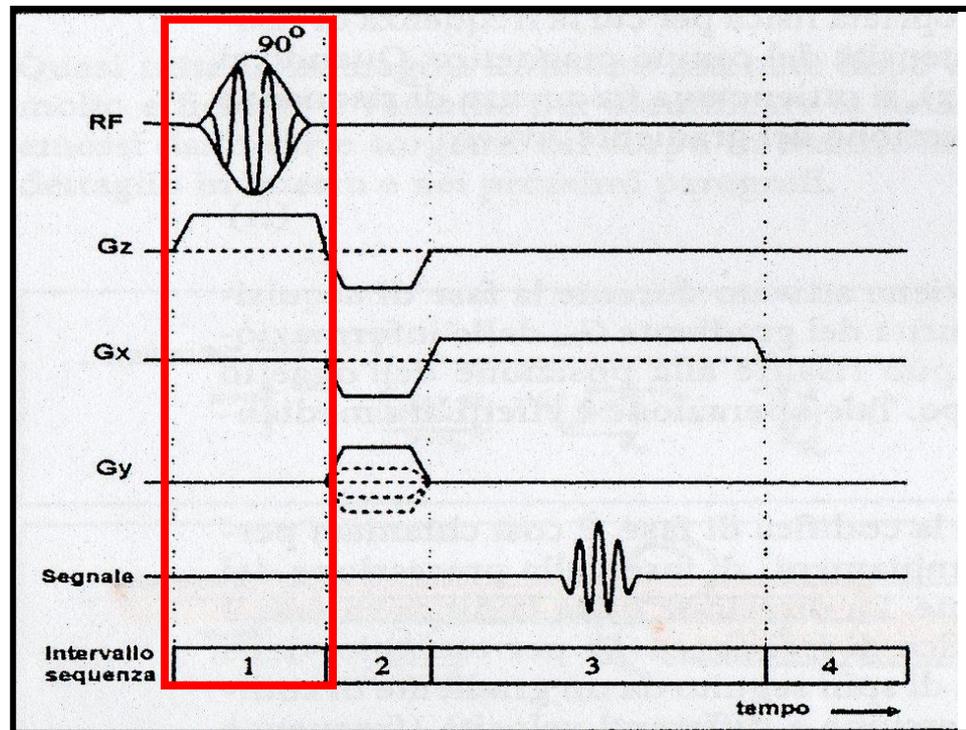


Image Localization

- **Phase encoding (G_{PE}) (y axis):** after the RF pulse and G_z , a short term magnetic field gradient is applied. This caused a phase offset proportional to the position along G_{PE} .

When G_{PE} is switch off, protons precess according to ω_0 , but their difference in phase allows to recognize their position.

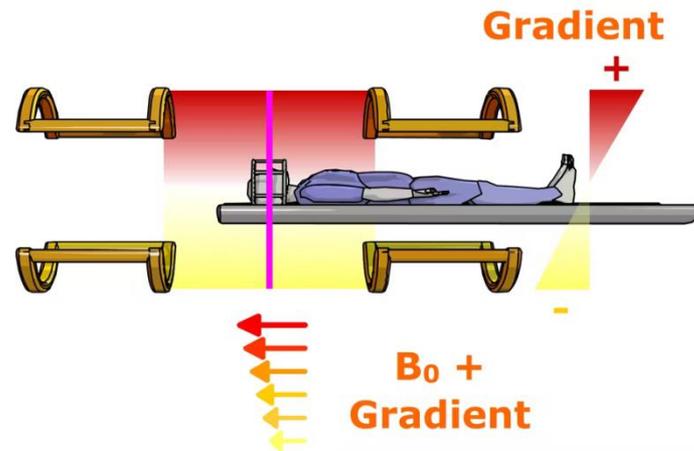


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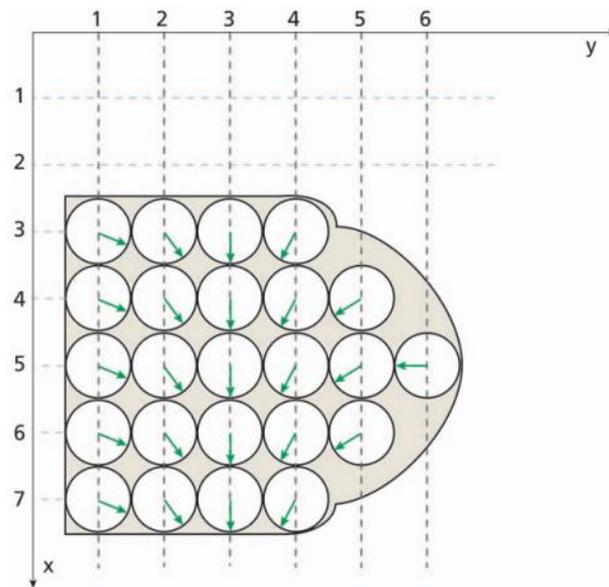


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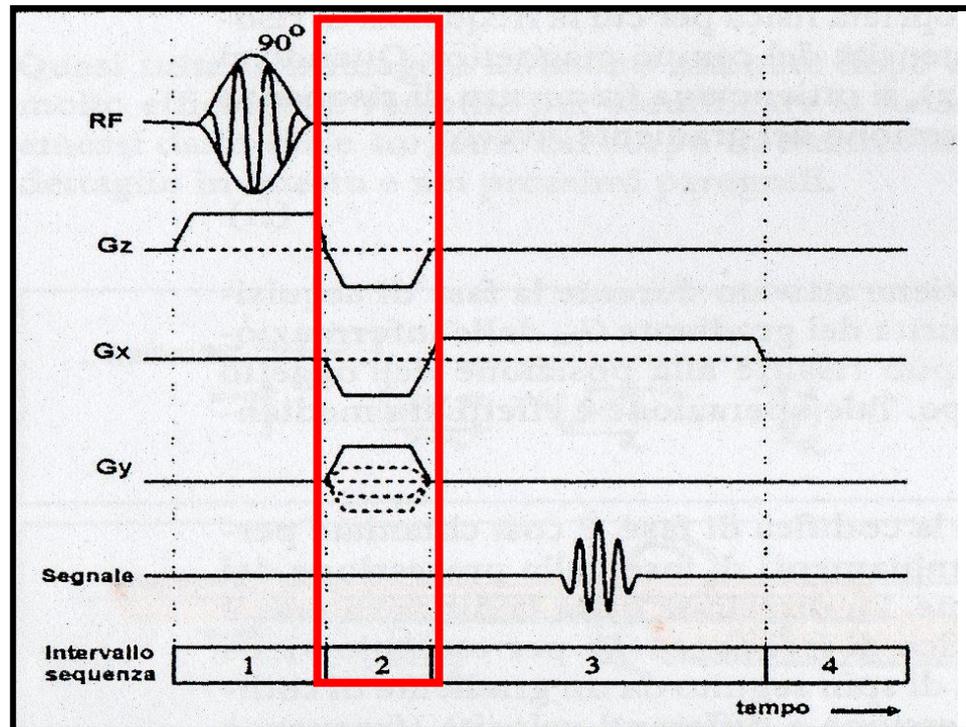


Image Localization

This sequence is repeated
with multiple
phase encoding steps
to allow a complete scan
along y axis.

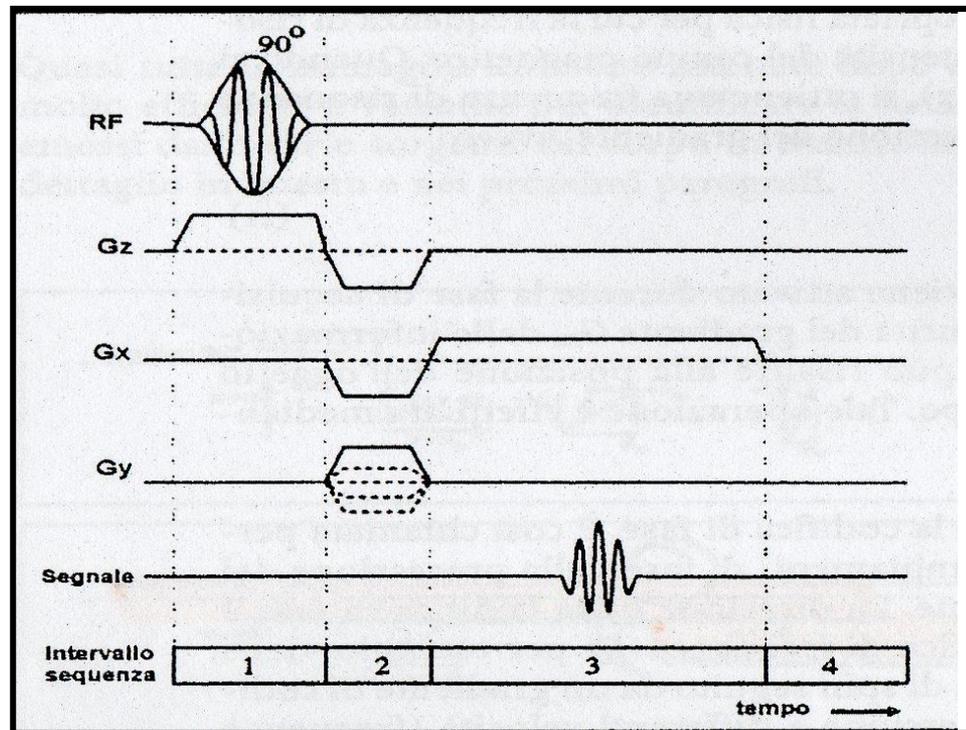


Image Localization

- **Frequency encoding (G_{FE}) (x axis):** is a magnetic field gradient delivered **during data reading**. The aim is to vary the resonant frequencies of spin across x-axis.
- The position along the frequency encoding gradient is become proportional to new precession frequency.

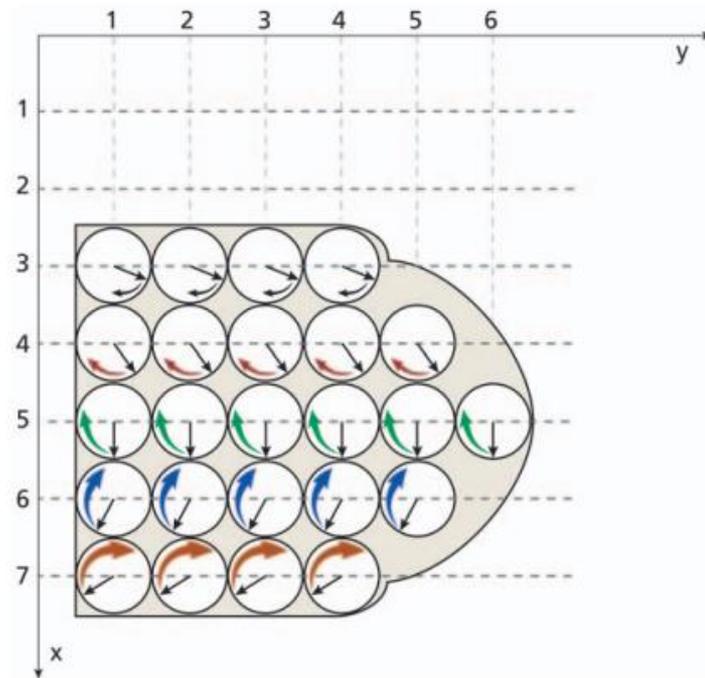


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- **Frequency encoding (G_{FE}) (x axis):** is a magnetic field gradient delivered **during data reading**. The aim is to vary the resonant frequencies of spin across x-axis.
- The position along the frequency encoding gradient is become proportional to new precession frequency.
- The final signal is the sum of all the frequencies.
- Finally, Furier transform converts the sum into each specific signal according to spatial position along x axis.

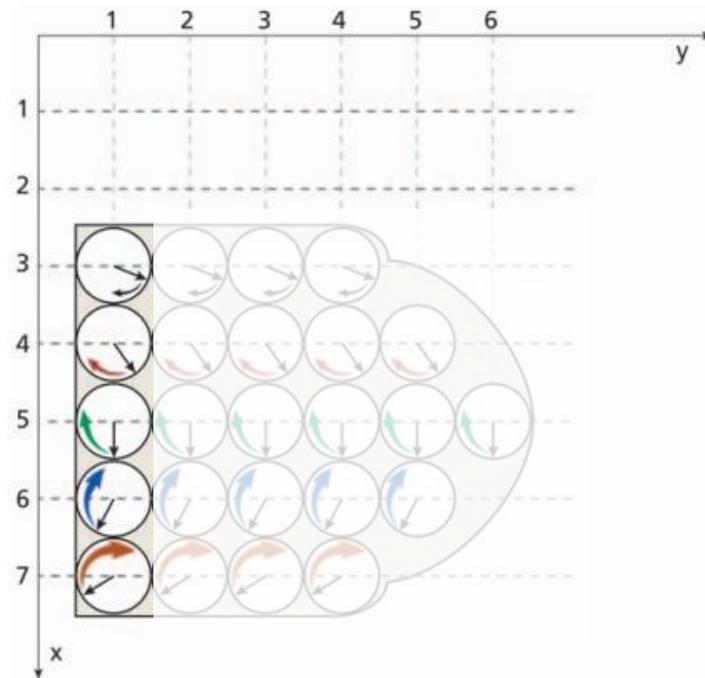


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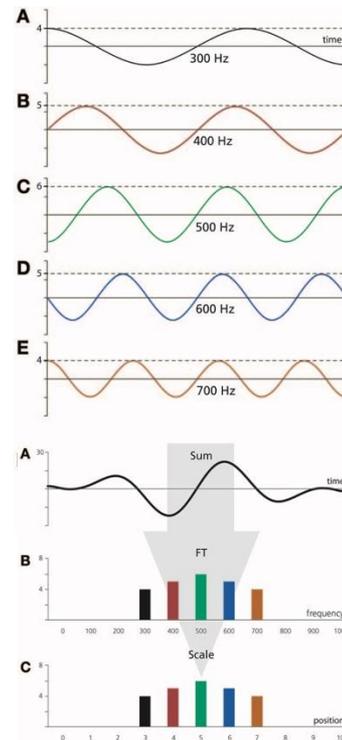
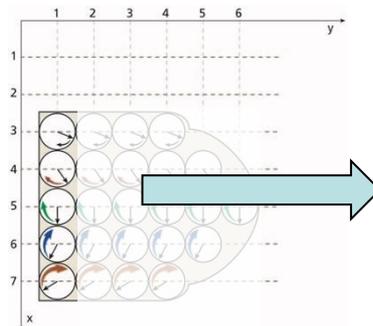
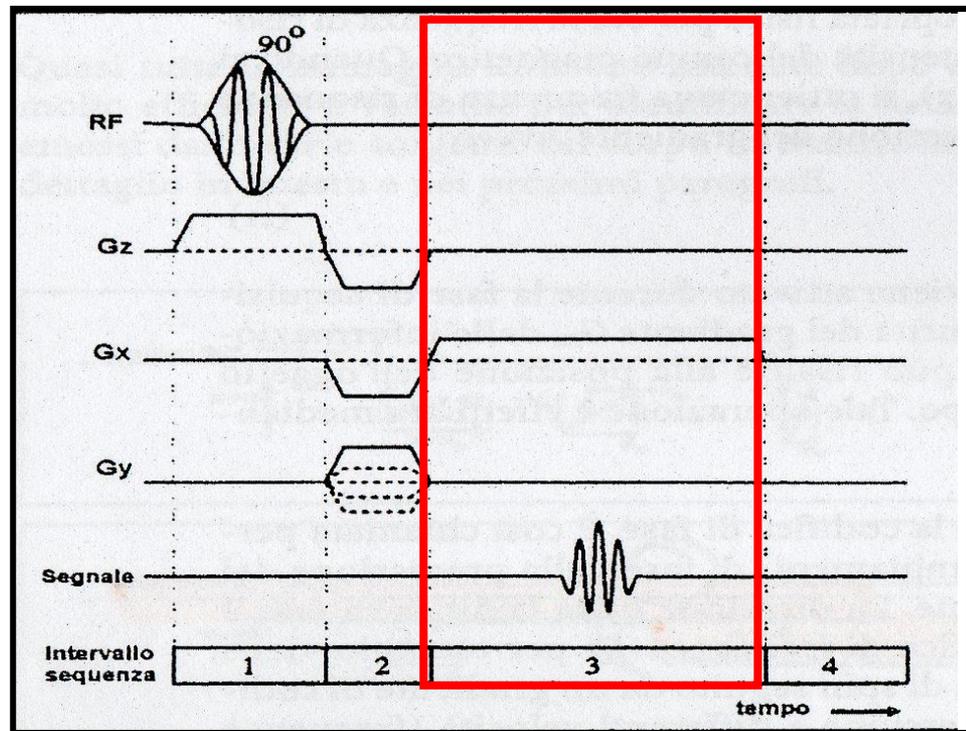
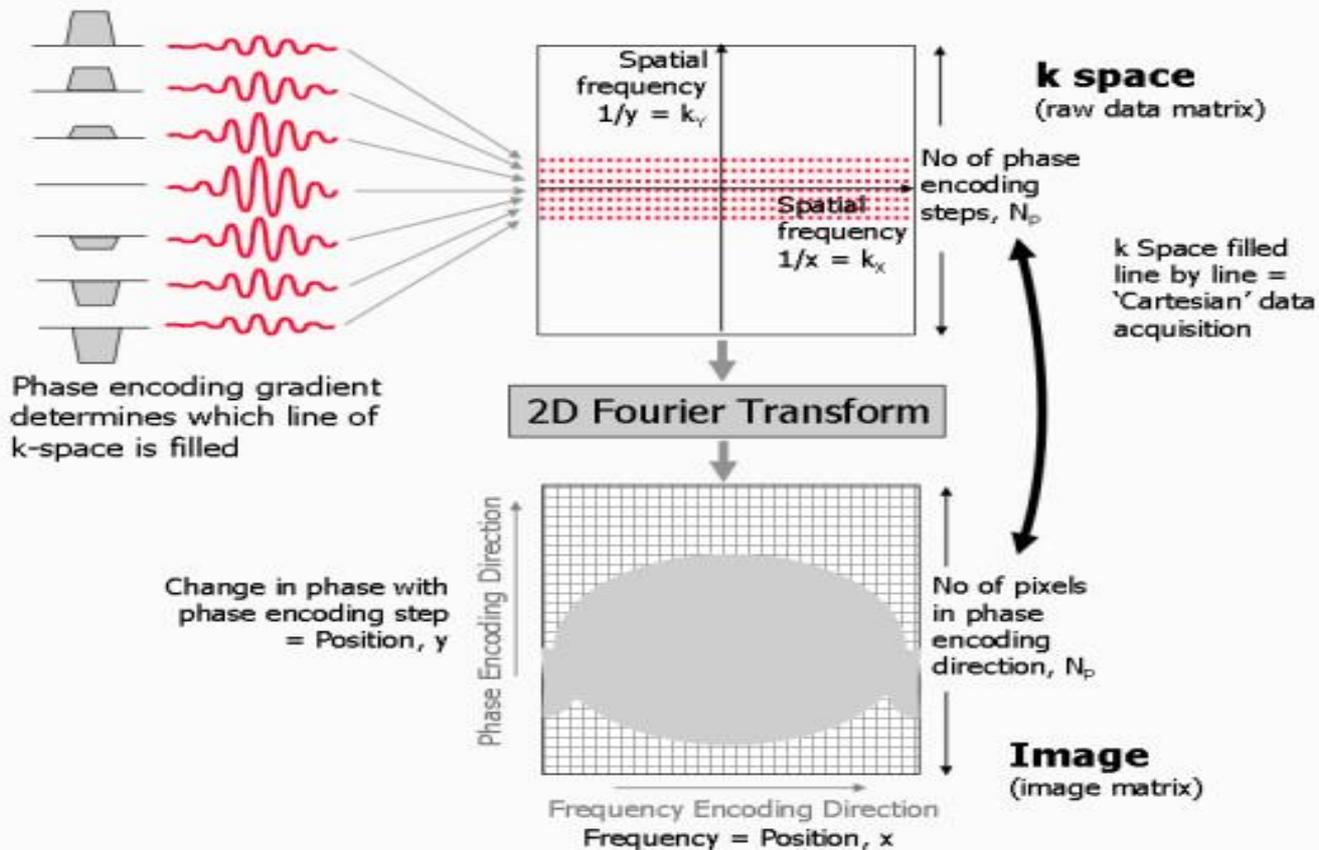


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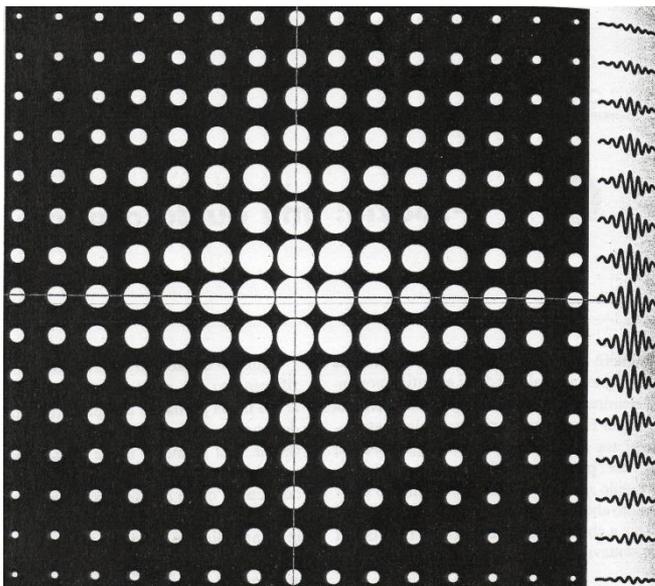


K-Space

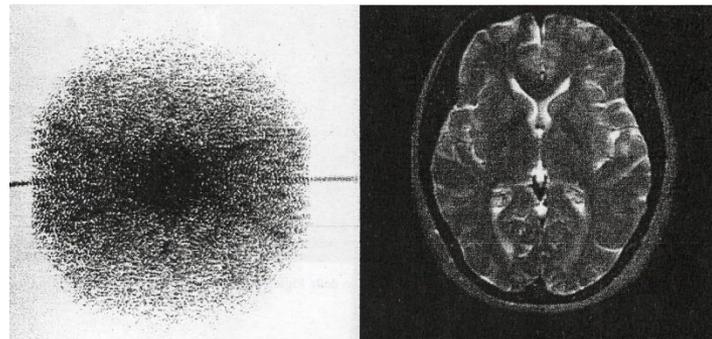
- K-space is the data storage where all information necessary to obtain a single MR image are recorded.
- In conventional imaging, the number of phase encoding steps (N_{PE}) corresponds to K-space rows, and to number of voxel in the y axis direction.
- The number of samples (NS) is the number of each signal is digitized into and represent the number of voxel in x axis.
- The Field of View is inversely related to the K-space

Frequency encoding

Phase
encoding



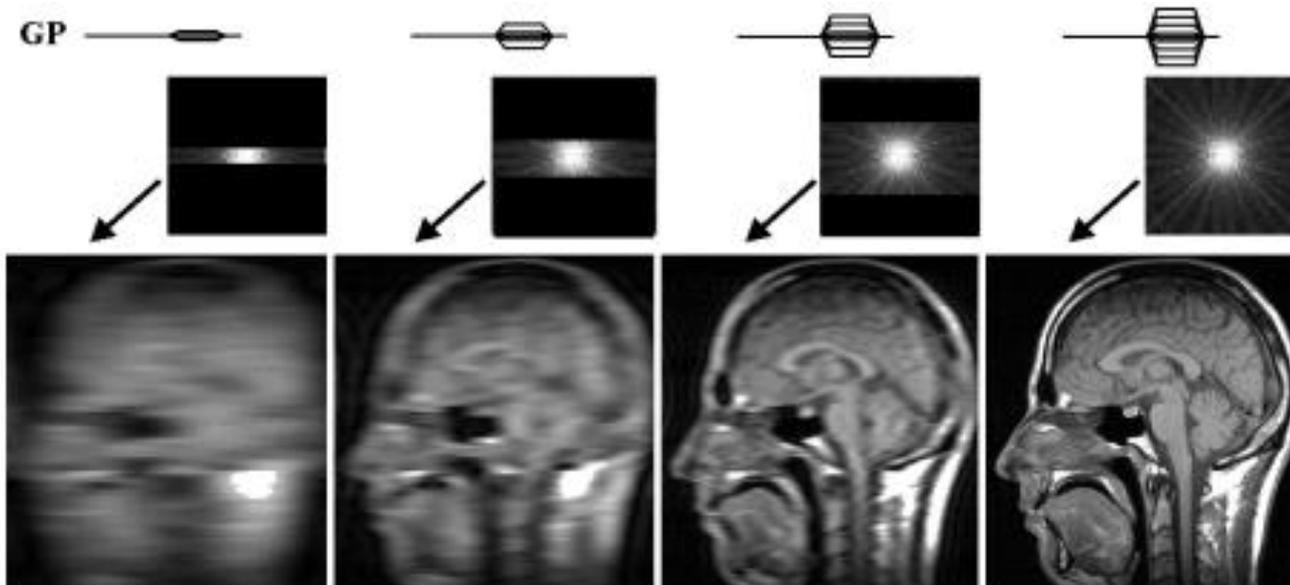
High spatial frequencies at the edges of K-space.



At centre there are low spatial frequencies, they are important for image contrast



K-Space





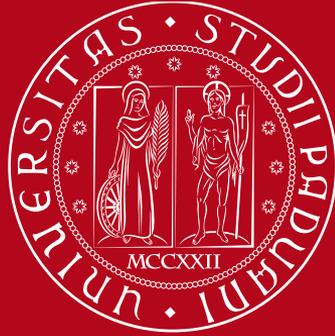
- Basic Physics
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2

parte

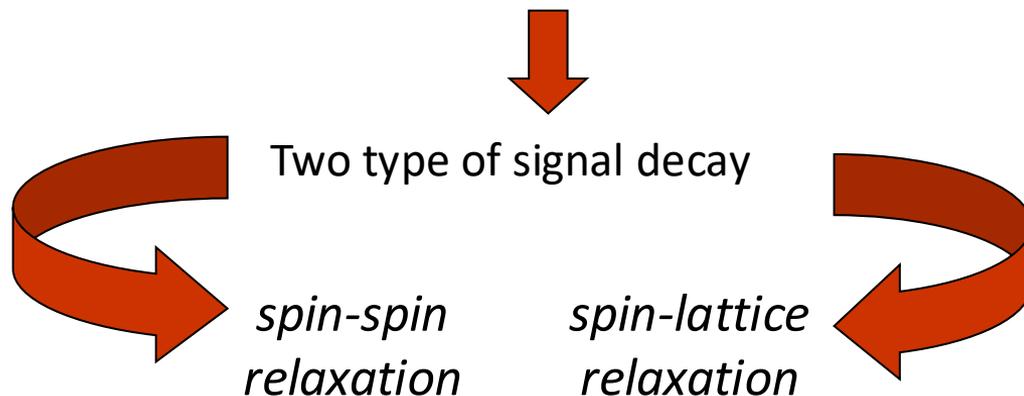


Relaxation times

Relaxation means return to equilibrium
retransmitting the RF energy.

If the nuclei behaved identically, then the RM imaging would provide only a map of nuclei concentration.

However additional mechanisms affect the releasing back energy that we measure to obtain images



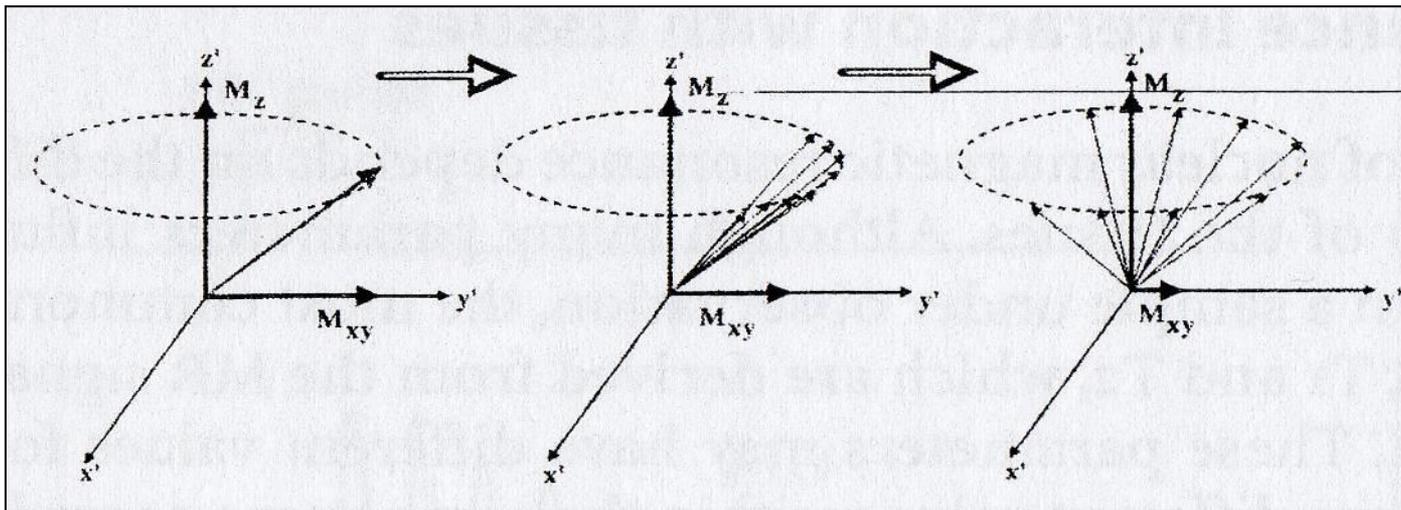
The aim is to achieve the lowest energy status, therefore the equilibrium (B_0).



Relaxation times

- Transverse Relaxation Time, T2
- Longitudinal Relaxation Time, T1
- Protonic Density, PD

Transverse magnetization decay

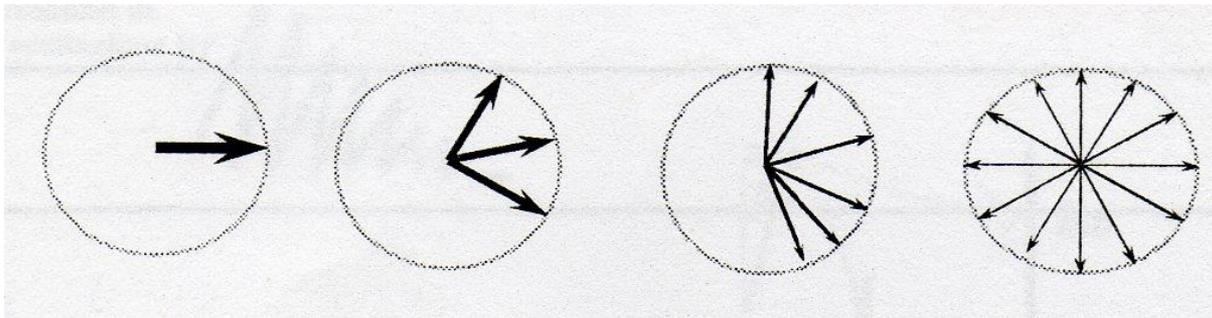


Transverse relaxation results from spins getting out of phase.

As spins move together, their magnetic fields interact randomly each-other causing a progressive loss in phase and therefore resulting in transverse magnetization decay.

Transverse magnetization decay

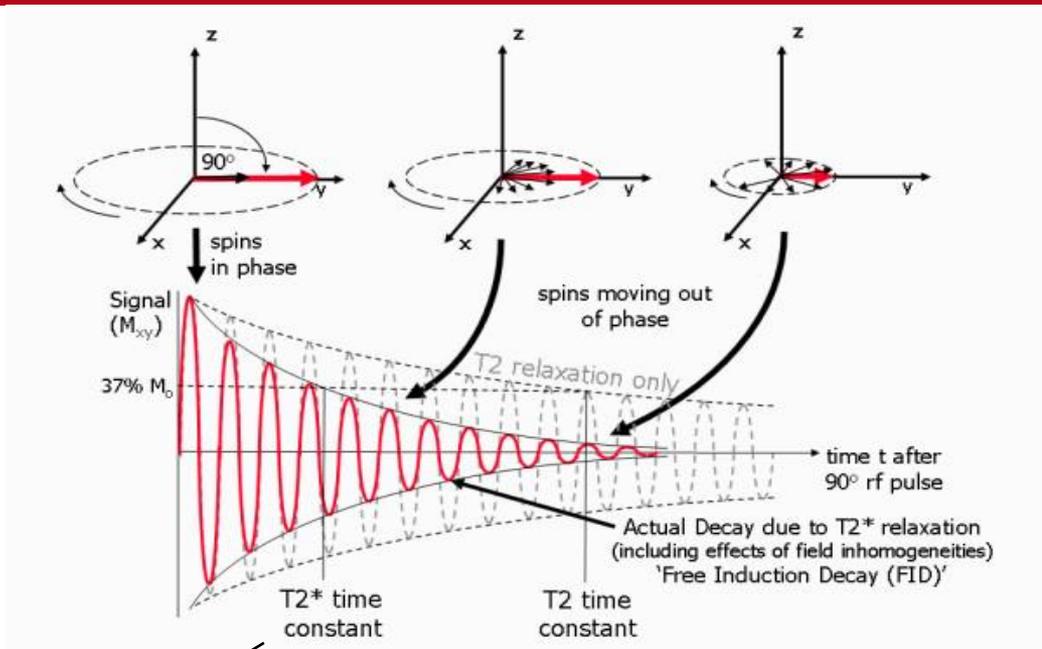
T2-time: the time when transverse magnetization has lost 63 % of its original value. It is tissue-specific.



Transverse relaxation results from spins getting out of phase.

As spins move together, their magnetic fields interact randomly each-other causing a progressive loss in phase and therefore resulting in transverse magnetization decay

Transverse magnetization decay



Static Magnetic Field inhomogeneities

Spin-Spin related signal loss corrected



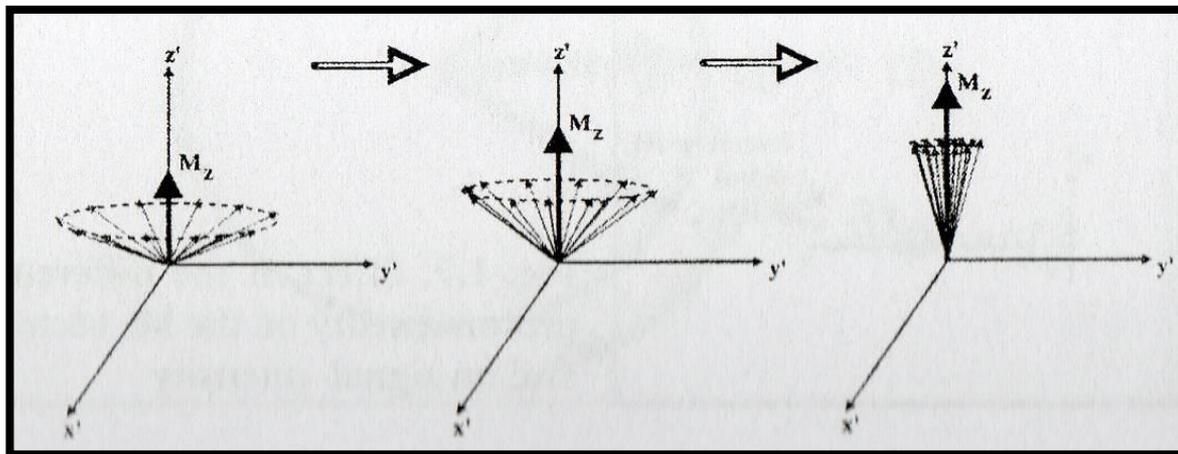
Relaxation times

- Transverse Relaxation Time, T2
- Longitudinal Relaxation Time, T1
- Protonic Density, PD

Longitudinal magnetization decay

Longitudinal relaxation is due to proton's energy loss.

Excited protons release their energy toward surrounding tissue (lattice).

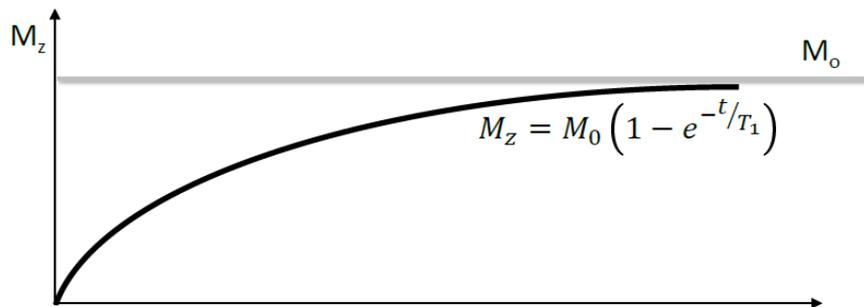
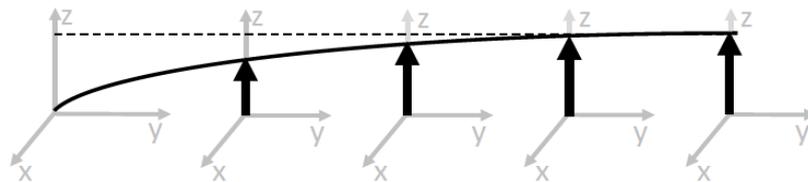




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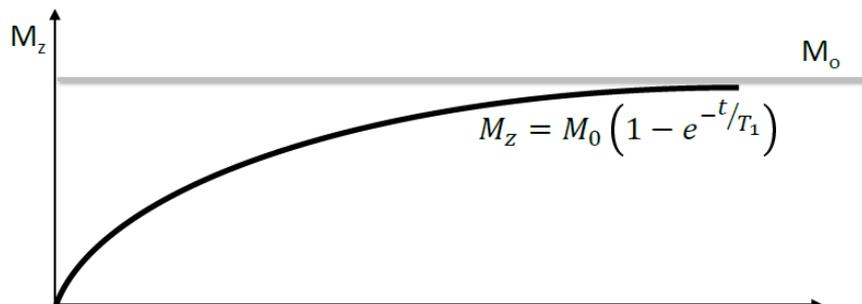
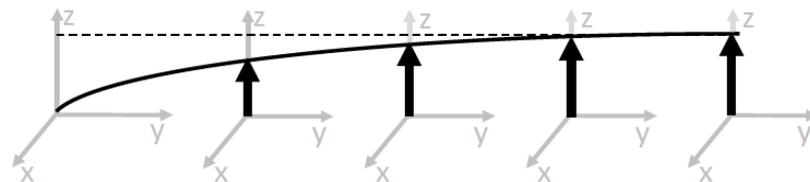
Excited protons release their energy toward surrounding tissue (lattice).





Longitudinal magnetization decay

T1-time: the time when longitudinal magnetization has returned to 63 % of its original value. It is tissue-specific. T1 increases with increasing B0 values





Each tissue, a couple of values

Tissue T2 and T1 values at $B_0 = 1.5T$

	T2 (msec)	T1 (msec)
Blood	362	895
Fat	108	192
Heart	45	800
Kidney	124	765

T2 time is always lower (or at least equal) than T1 time.



Table 7.7. Characteristics of the MRI signal from different tissues in the T1- and T2-weighted images

Tissue	T1	T2
Liquid	Low (—)	High (+++)
Myxoid	Low (—)	High (+++)
Collagen	Low (—)	Low/High (—/++) *
Adipose	High (+++)	High (++)
Necrosis	Low (—)	High (+++)
Fibrous	Low (—)	Low/High (—/++) *
Calcium	Low (—)	Low (—)
Vascularized	Low (—)	High (++)

* The type of signal and the capture depends on the vascularization and the cellularity of such tissues



Table 11.1. MR Signal characteristics of Hemoglobin catabolites

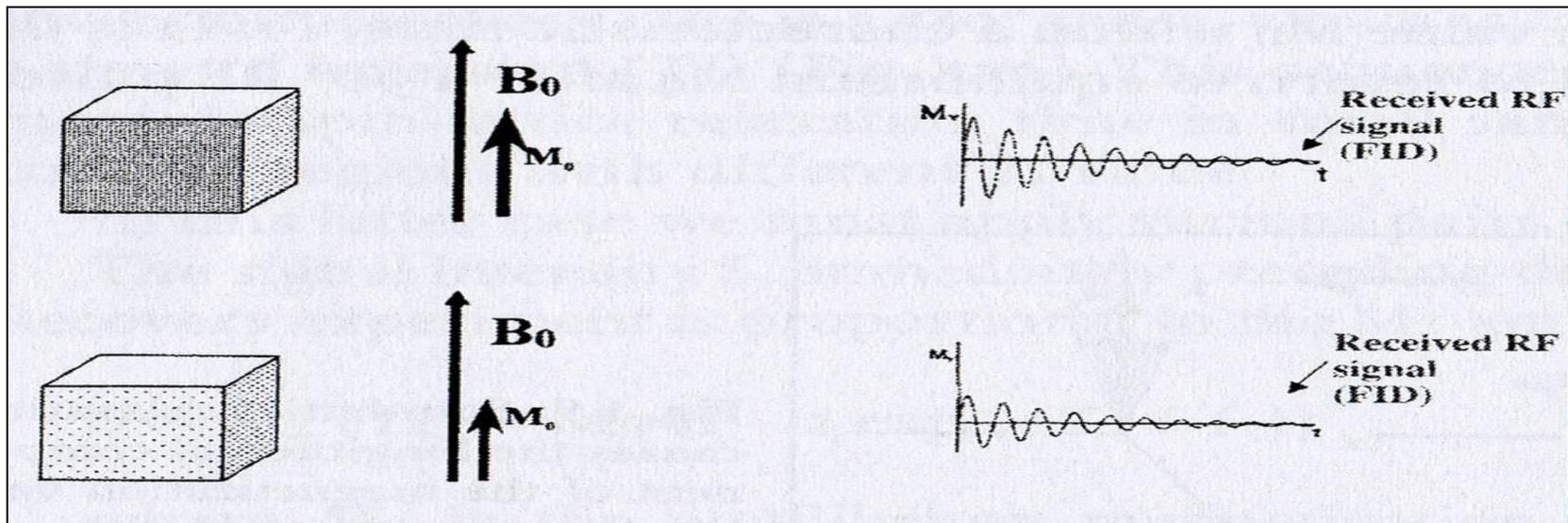
Hemoglobin (Hb) catabolites	T₁	T₂
Oxy-Hb hyper acute phase (0-12 hours)	hypo	hyper
Deoxy-Hb acute phase (12-72 hours)	hypo	hypo
Intracellular met-Hb early sub-acute phase (3-5 days)	hyper	hypo
Intracellular met-Hb late sub-acute phase (7-15 days)	hyper	hyper
Hemosiderin chronic phase (15 days)	hypo	hypo



Relaxation times

- Transverse Relaxation Time, T2
- Longitudinal Relaxation Time, T1
- Protonic Density, PD

Protons Density



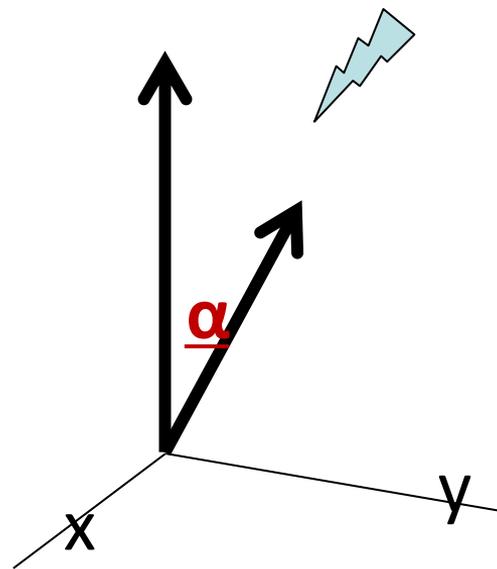
Higher protons density \rightarrow Higher $M_0 \rightarrow$ Higher FID



- ▶ FA (flip angle): RF delivery time
- ▶ TR (repetition time): amount of time between excitation pulses.
- ▶ TE (echo time): amount of time between excitation and data acquisition/read out

$$\alpha = \gamma B_1 t$$

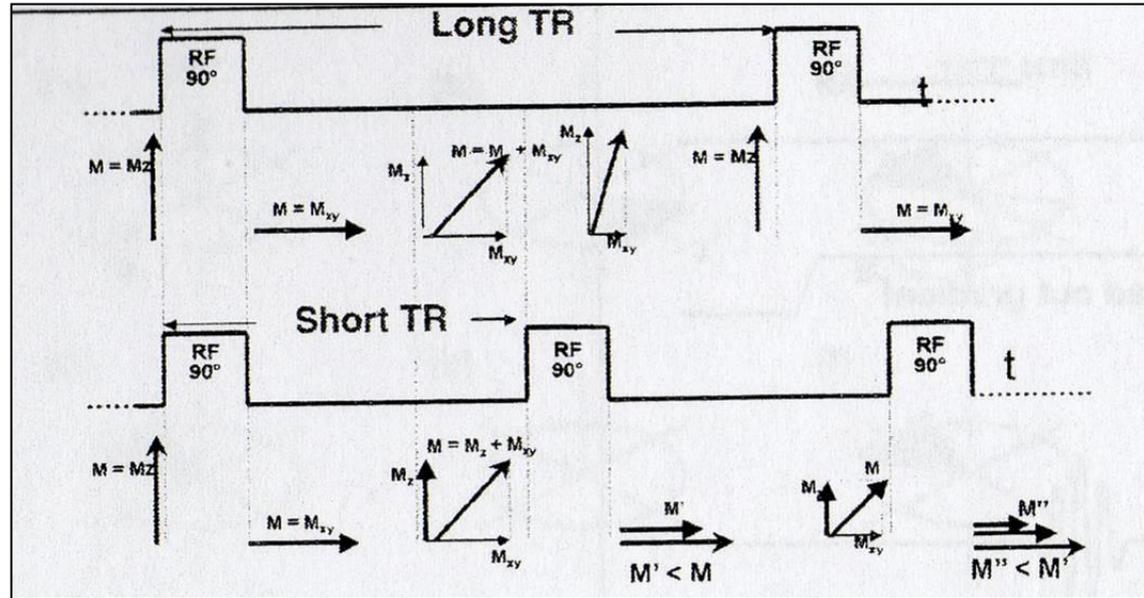
- In order to reduce imaging times, a FA less than 90° is set.
- The RF pulse only tips a component of M into Mxy plane, which means lower signal.
- But remaining a residual component of magnetisation in Mz, that allows to start with a new RF pulse without waiting a complete Mz restoration and without signal saturation.



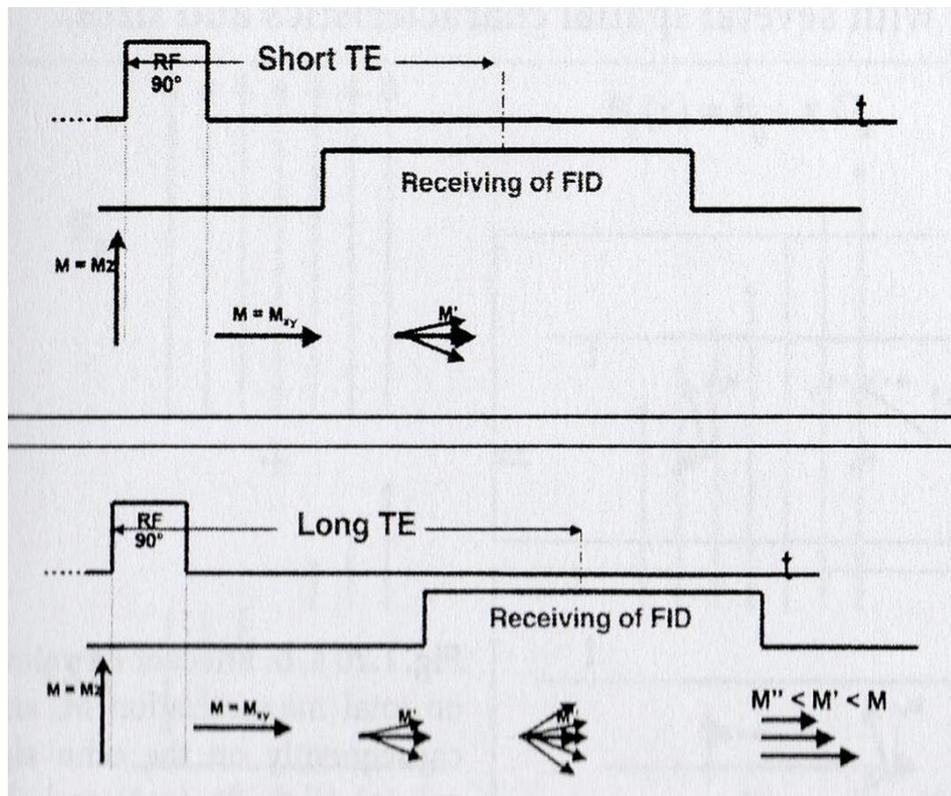
TR – repetition time

- **TR long:**
- Longer scan
- Higher signal (high SNR).

- **TR short**
- Reduce scan duration;
- Signal saturation;
- Progressive reduction of signal



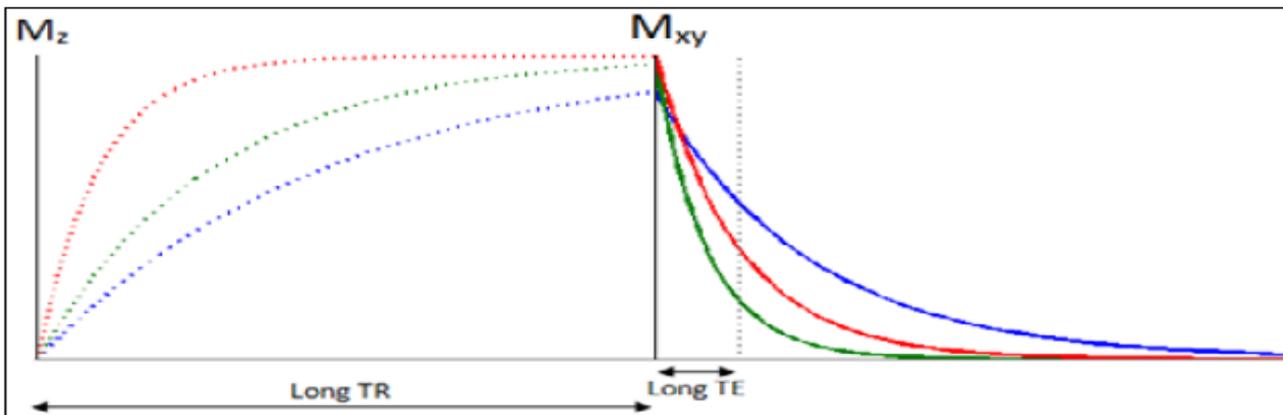
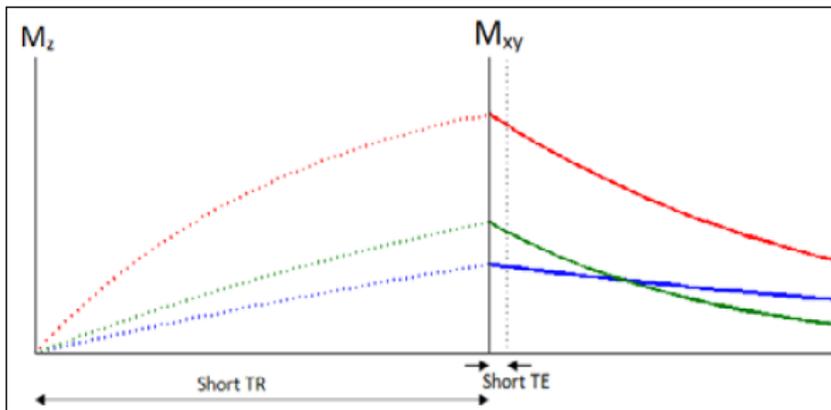
TE – echo time



Short TE minimises
differences in
transverse
magnetisation decay

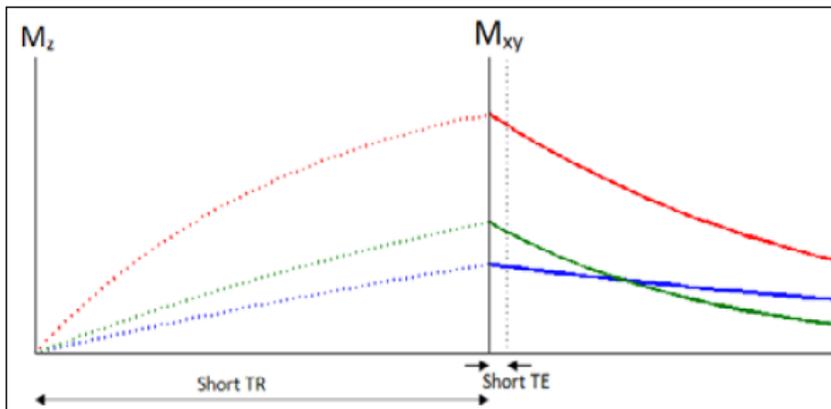


T1-w; T2-w; DP-w

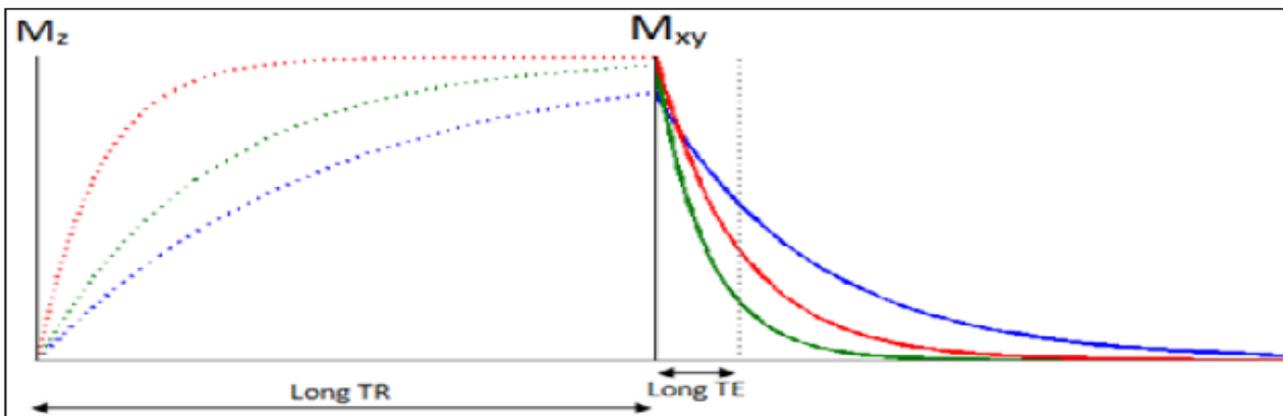




T1-w; T2-w; DP-w

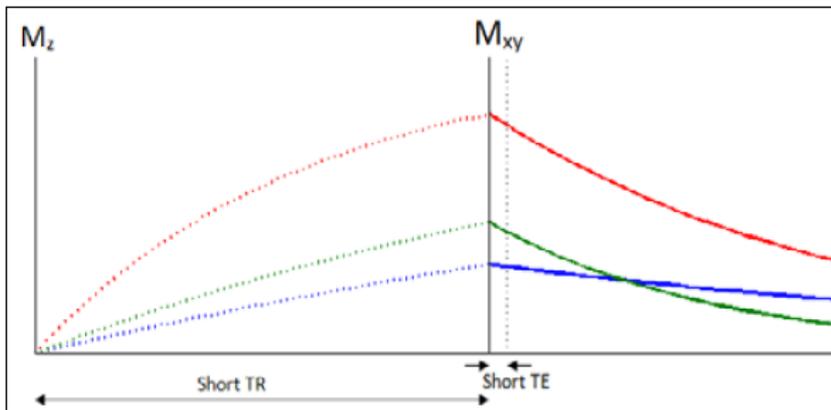


T1-w

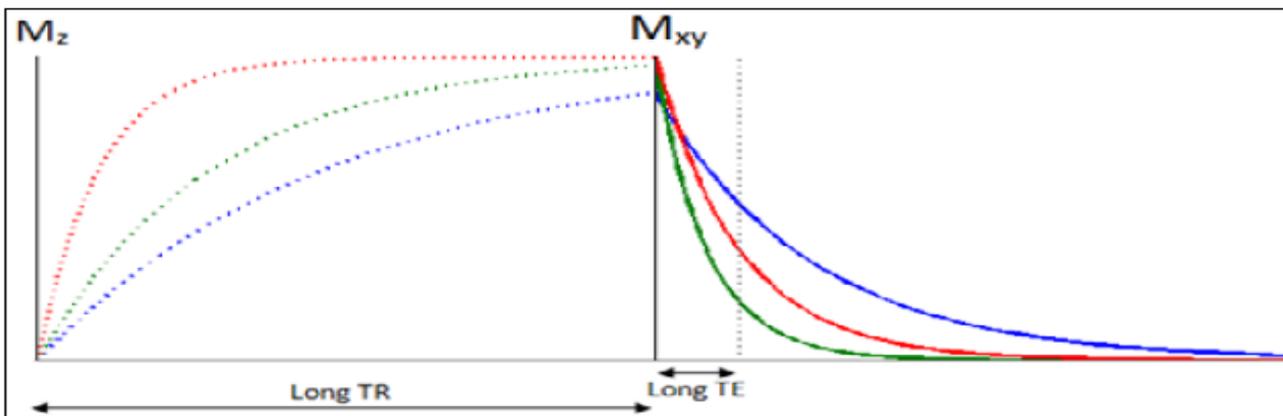




T1-w; T2-w; DP-w



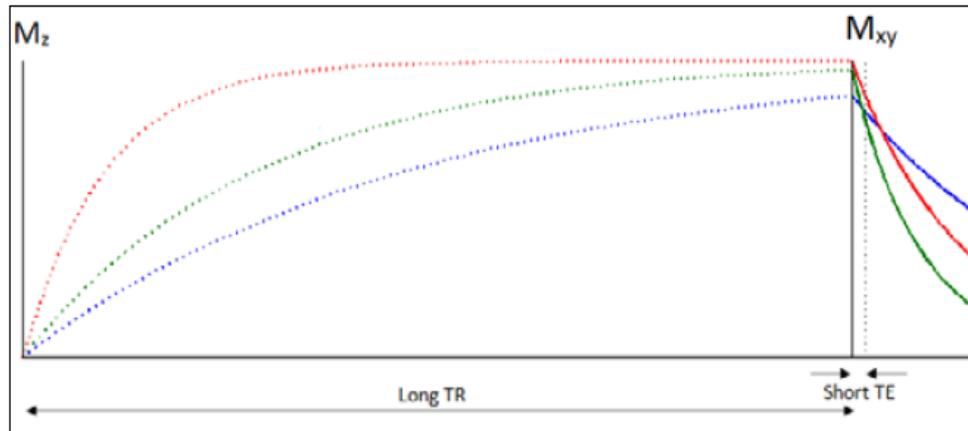
T1-w



T2-w

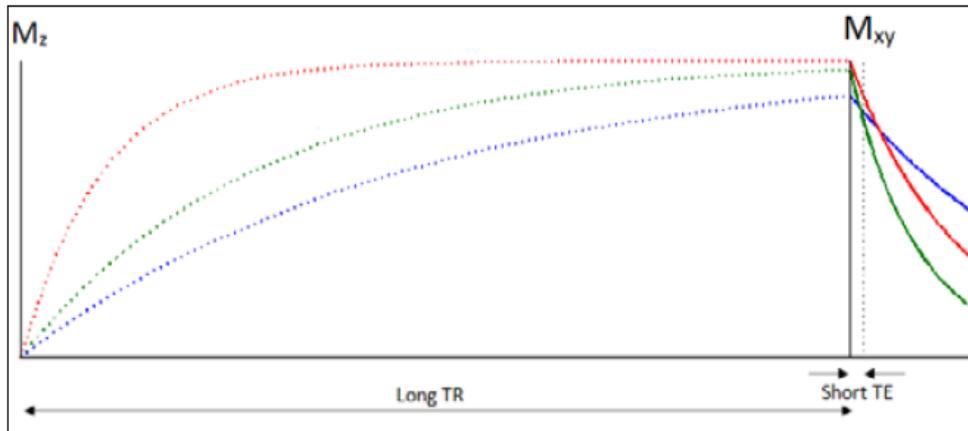


T1-w; T2-w; DP-w





T1-w; T2-w; DP-w



DP-w

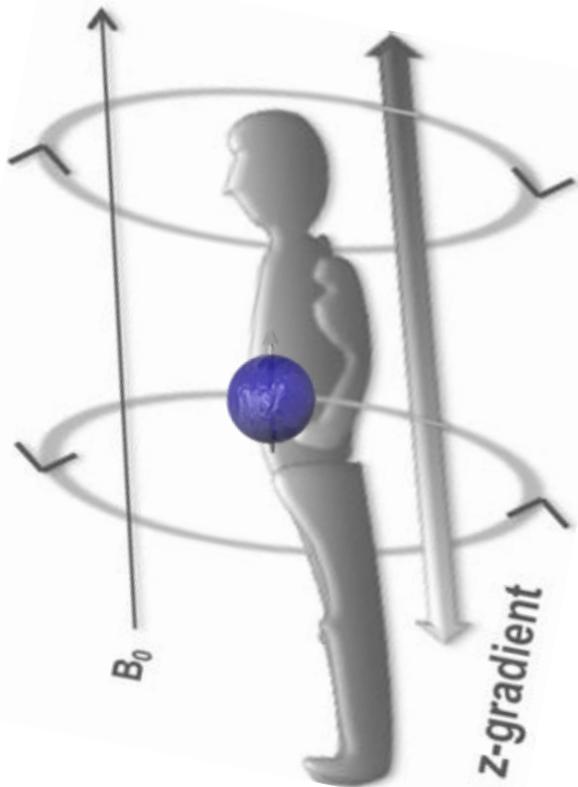


T1-w; T2-w; DP-w

Weighting	Short TR	Long TR
Short TE	T1	PD
Long TE	-	T2/T2*

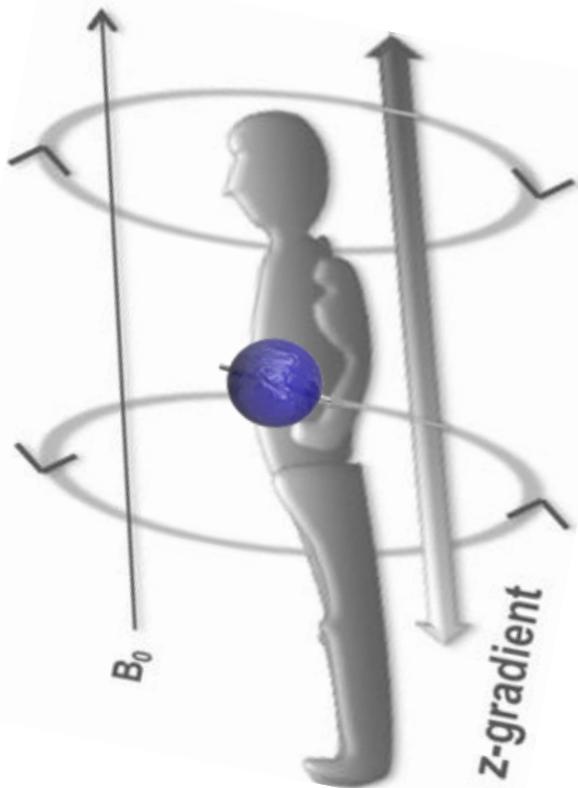


Make RM easy

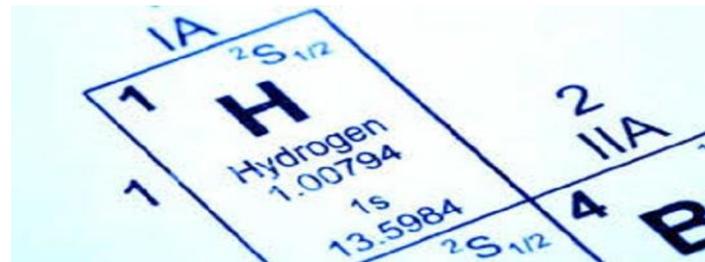




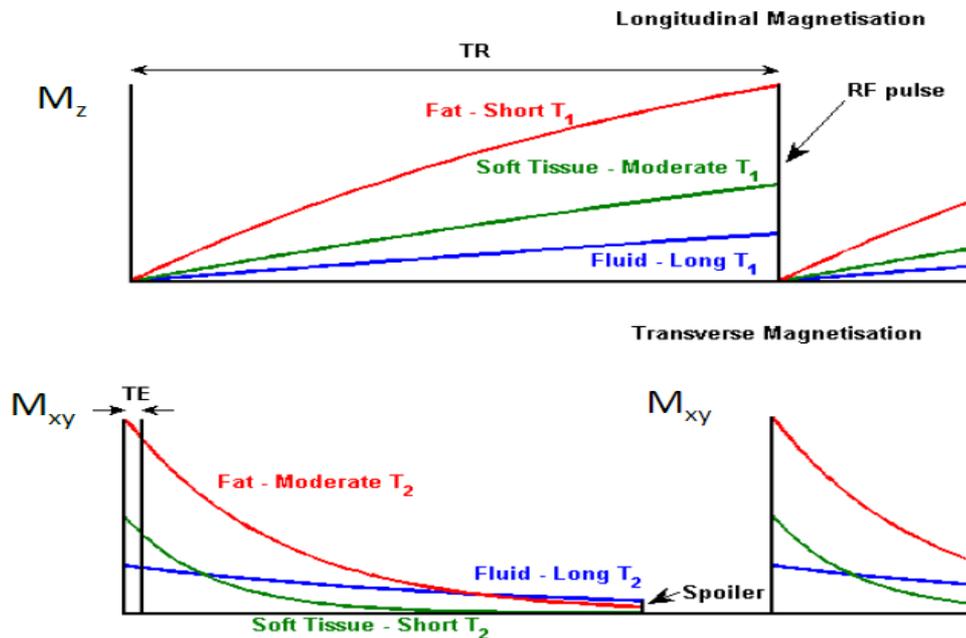
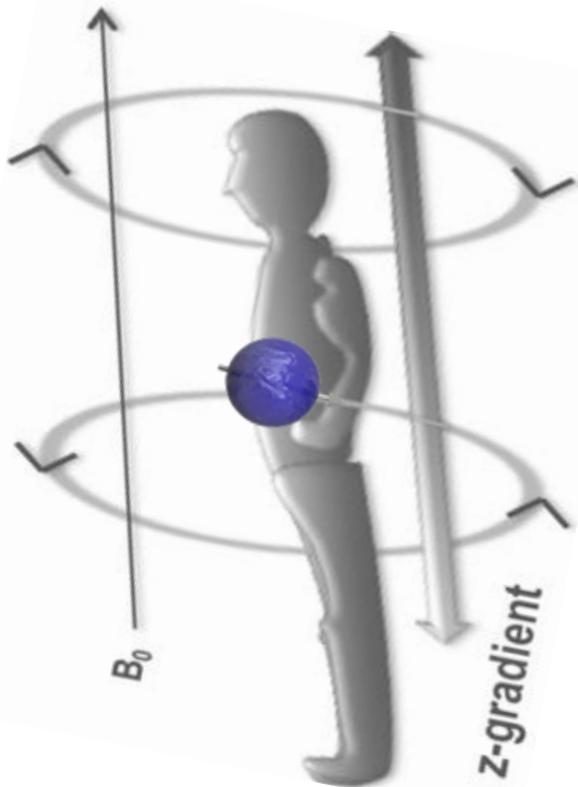
Make RM easy



RF specific ^1H impulse



Make RM easy





- Basic Physics
- The creation of MR signal
- The acquisition of the signal
- Image location
- Image contrast

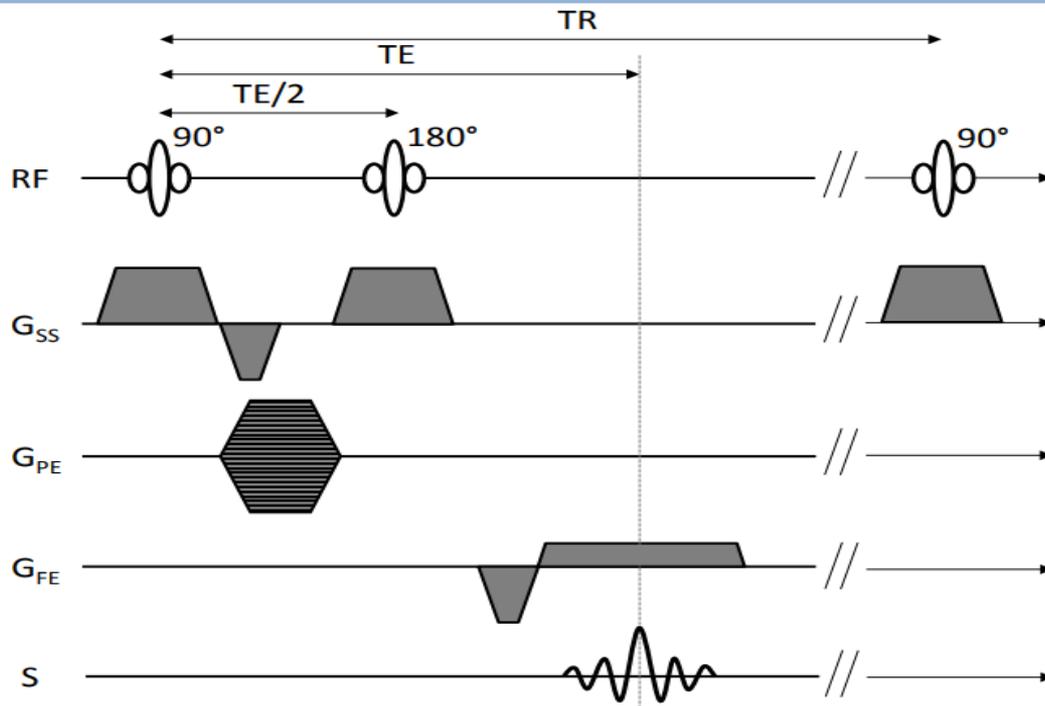


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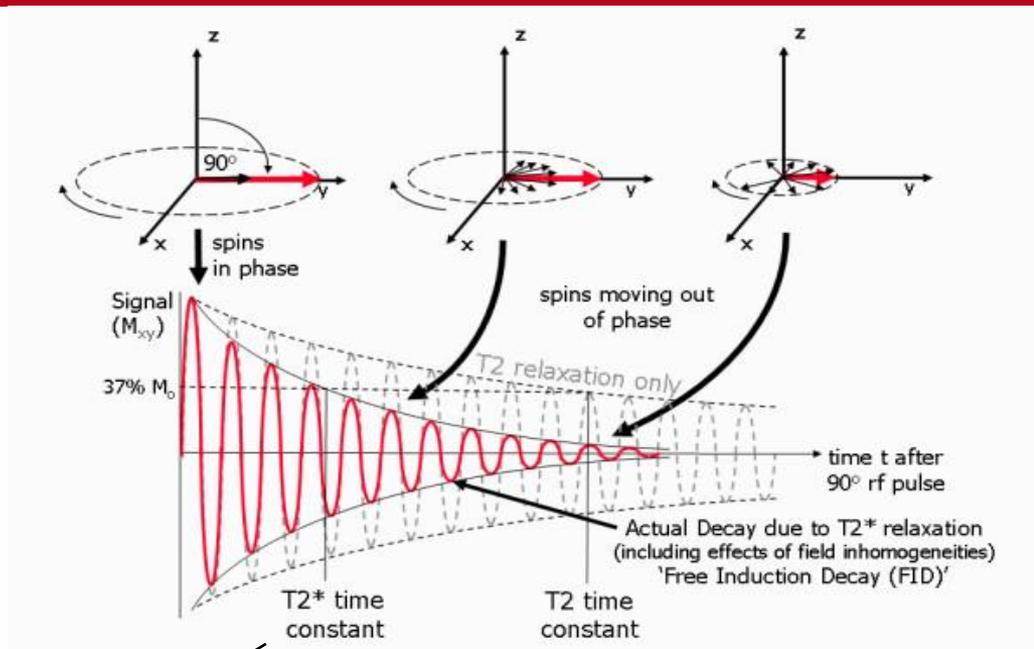


- ▶ SPIN-ECHO (SE)
- ▶ GRADIENT-ECHO (GRE)
- ▶ INVERSION-RECOVERY (IR)
- ▶ CARDIAC SYNCR

SE Pulse Sequence Diagram



The base sequence consists in a 90° pulse followed by one of 180° . The time between the two pulses is $TE/2$.

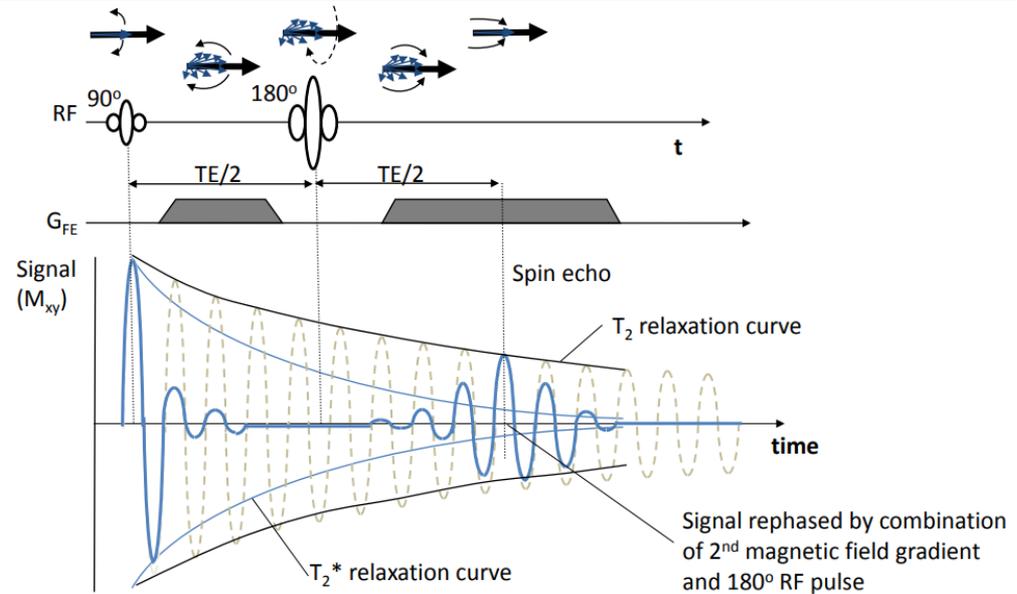


Static Magnetic Field inhomogeneities

Spin-Spin related signal loss corrected

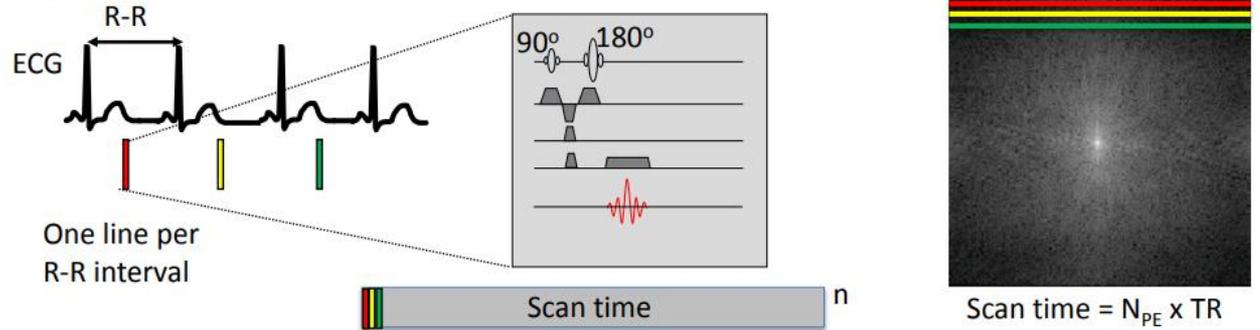
- The **refocusing pulse** changes the sign of the relative phase in the xy plane
- The spins continue to gain or lose phase in the same direction as before by virtue of the static field inhomogeneities
- At the TE time all spins come back into phase again
- Throughout the spin echo sequence the irreversible spin-spin interaction continues so that the measured signal is still affected by T2 dephasing
- Increased SNR
- Tissue specific

Spin Echo Formation



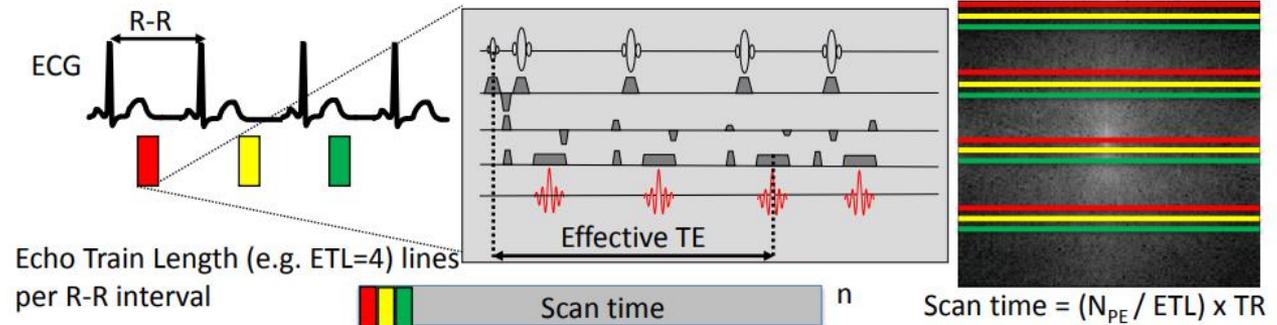
Turbo Spin-Echo

Conventional Spin Echo



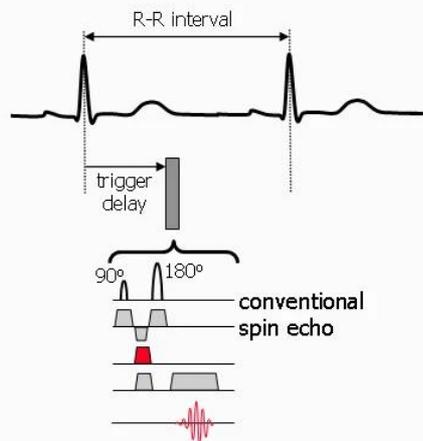
Fast/Turbo Spin Echo

- The transverse magnetisation is repeatedly refocused using consecutive 180 pulses
- More than one line are acquired after each 90 pulse, **echo train length**

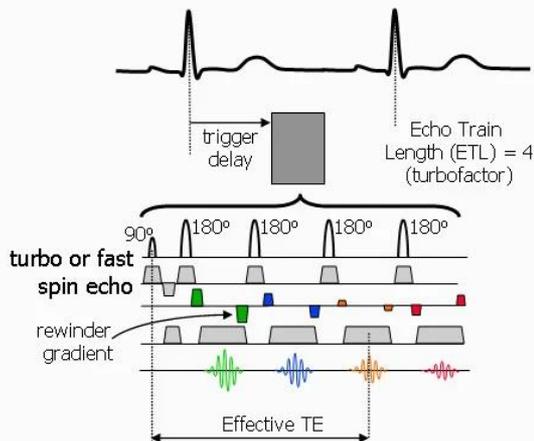
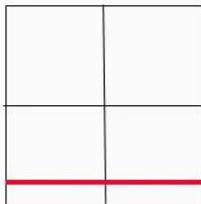




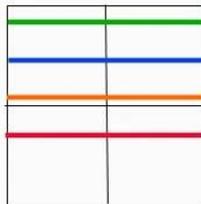
Turbo Spin-Echo



One line of
k-space filled
per R-R interval



multiple lines (4)
of k-space filled
per R-R interval



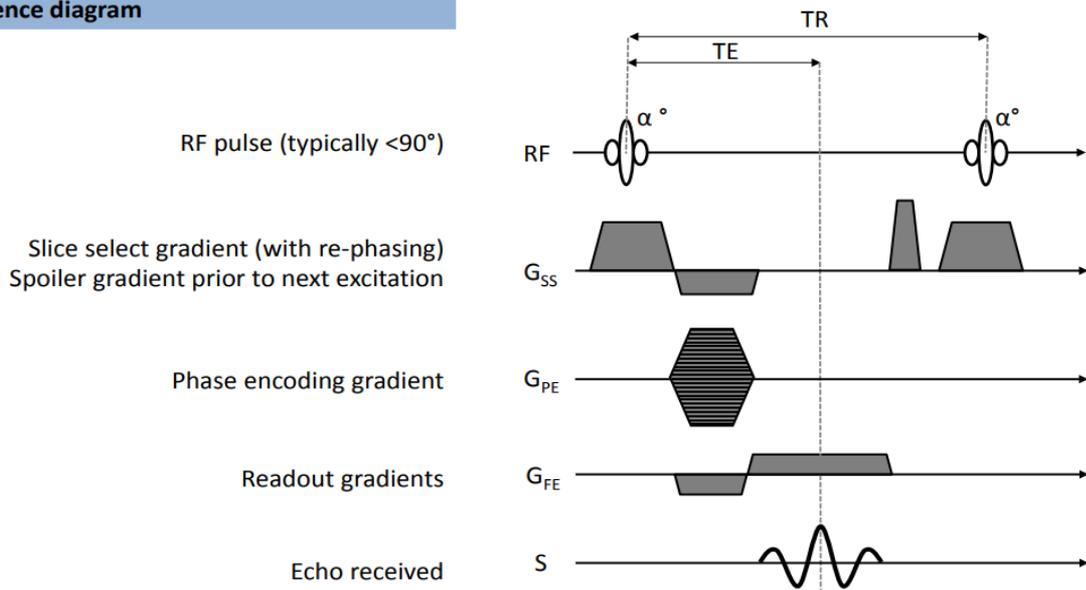
Multiple lines of k-space are filled within each R-R interval by applying a phase encoding gradient with a different amplitude to each echo

Gradient-Echo

The consequence of a **low-flip** angle excitation is a faster recovery of longitudinal magnetization that allows shorter TR/TE and decreases scan time.

A bipolar **readout gradient** (which is the same as the frequency-encoding gradient) is required to create an echo. The gradient echo formation results from applying a dephasing gradient before the frequency-encoding or readout gradient.

Pulse sequence diagram



Gradient-Echo

The consequence of a **low-flip** angle excitation is a faster recovery of longitudinal magnetization that allows shorter TR/TE and decreases scan time.

A bipolar **readout gradient** (which is the same as the frequency-encoding gradient) is required to create an echo. The gradient echo formation results from applying a dephasing gradient before the frequency-encoding or readout gradient.

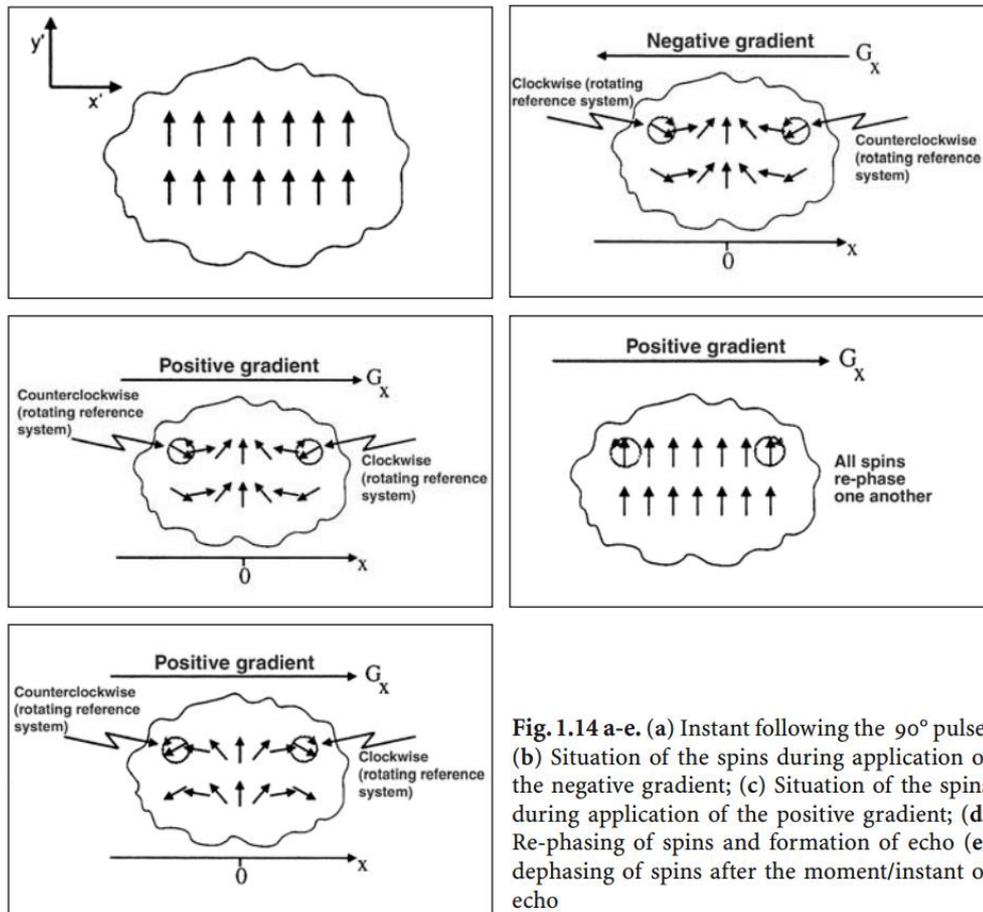
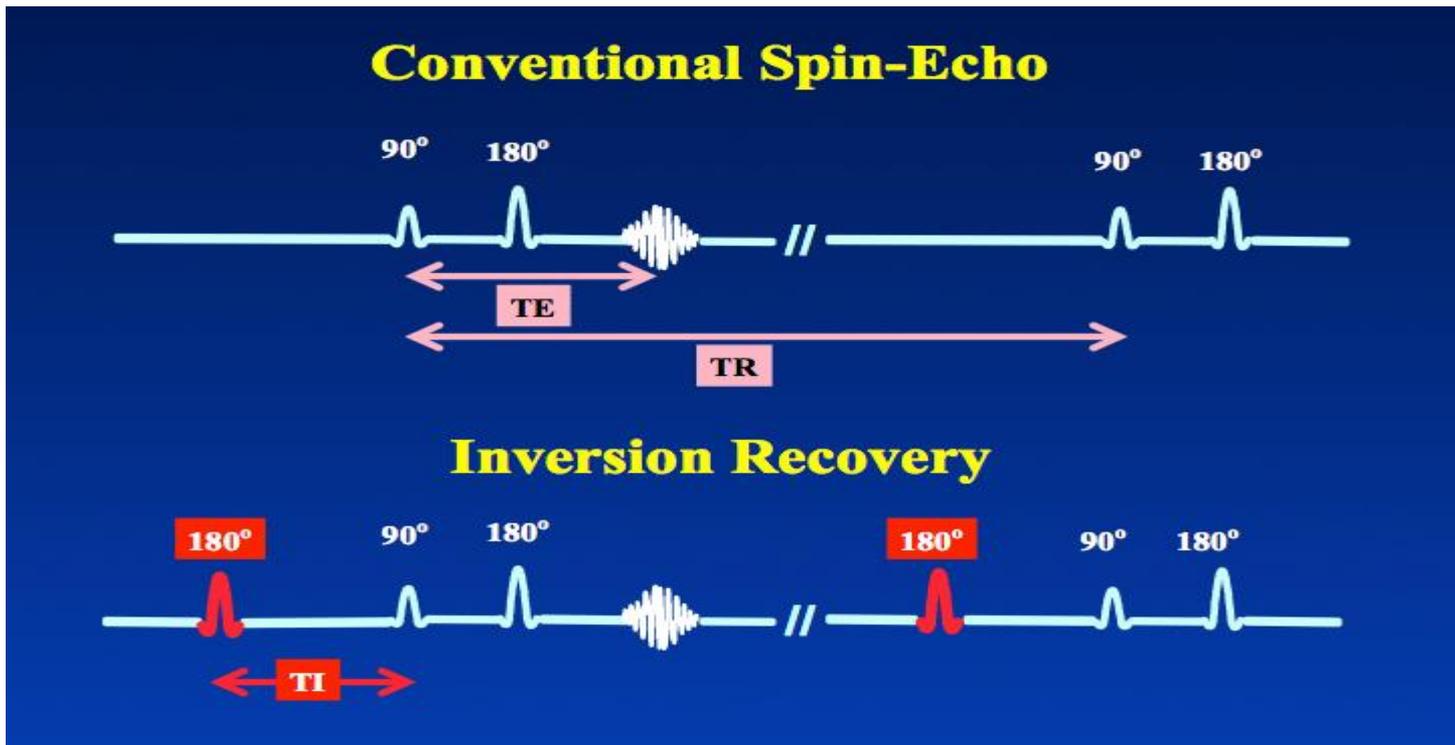


Fig. 1.14 a-e. (a) Instant following the 90° pulse; (b) Situation of the spins during application of the negative gradient; (c) Situation of the spins during application of the positive gradient; (d) Re-phasing of spins and formation of echo (e) dephasing of spins after the moment/instant of echo

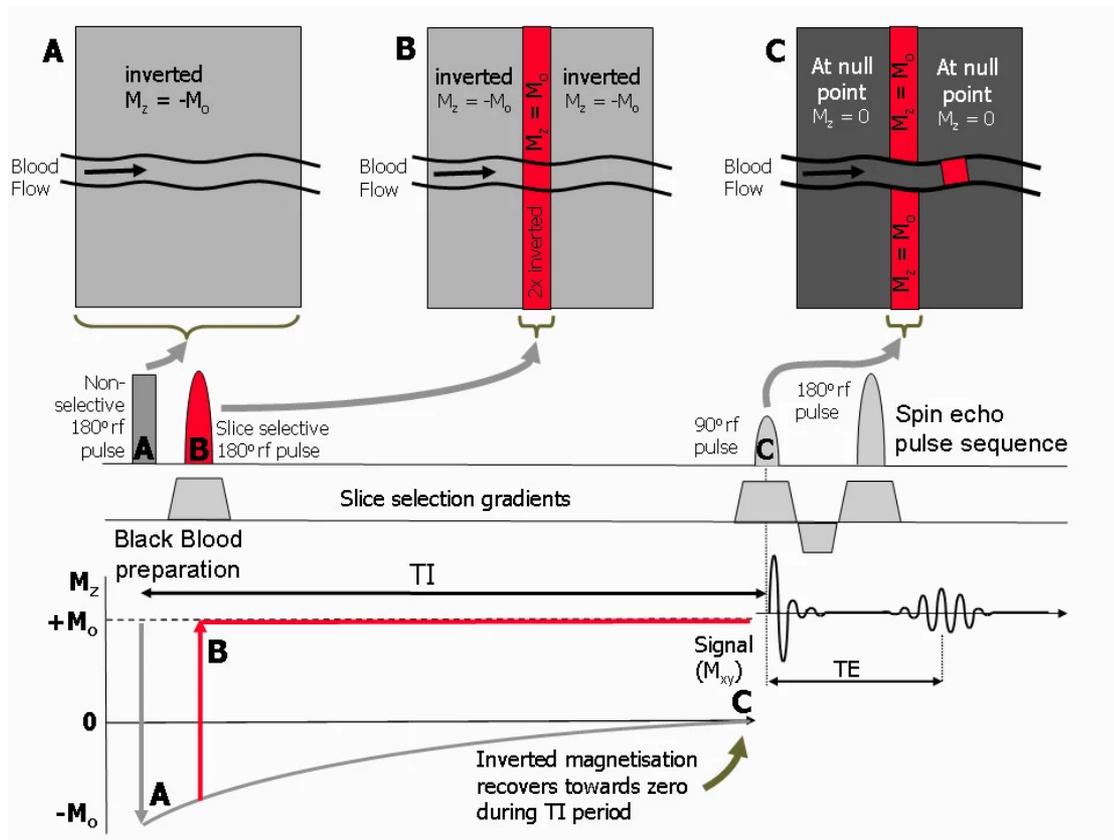


- Gradient echo sequences with low flip angles
- Short TR
- The remnant XY magnetization is not spoiled but maintained using additional rephasing and dephasing gradients (T1/T2w)
- High SNR but good shimming required
- Low flow sensitivity

Inversion-Recovery



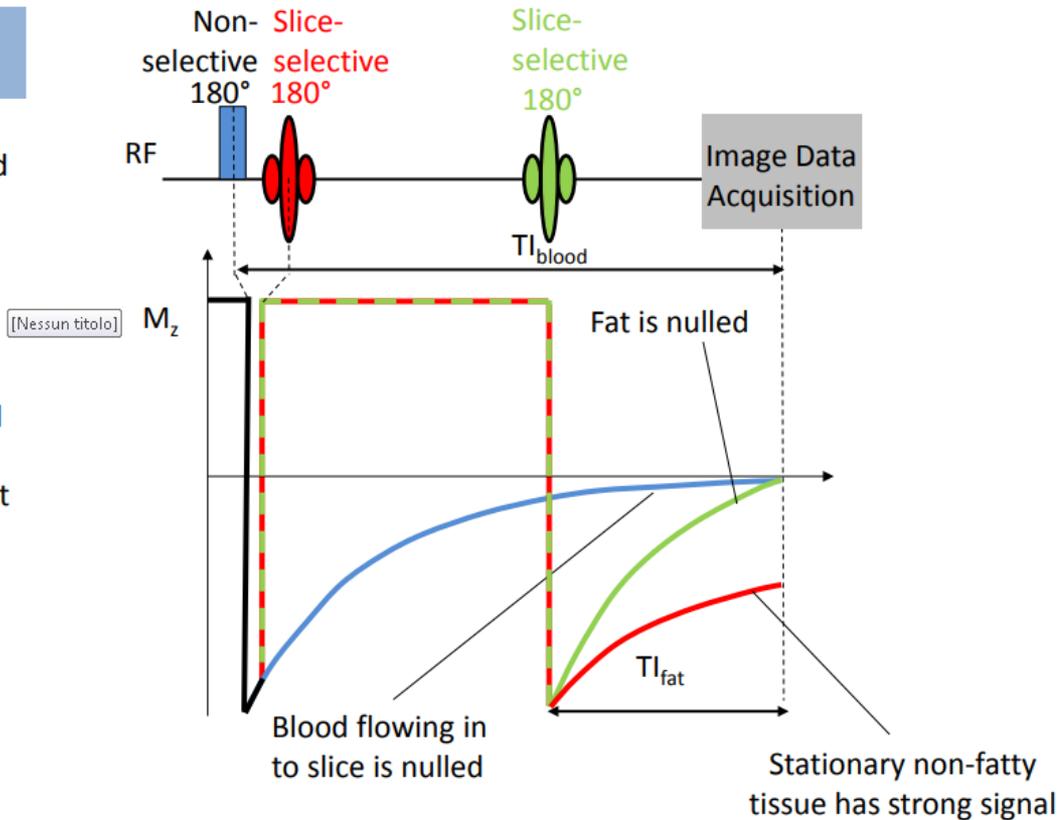
Black Blood



Inversion-Recovery

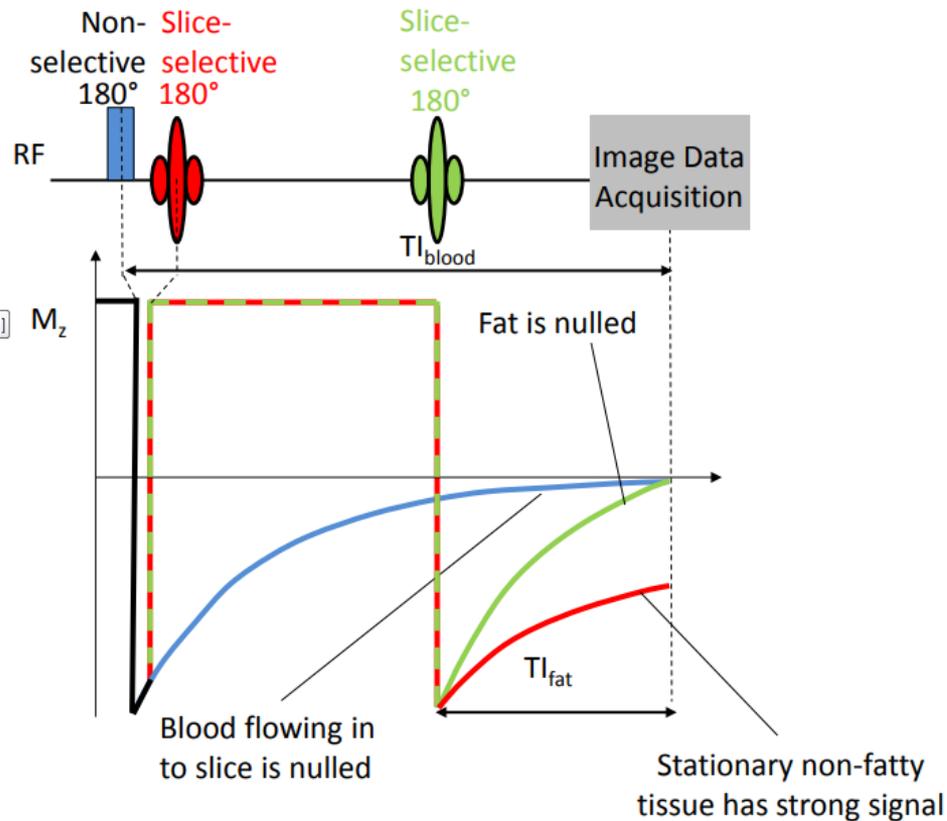
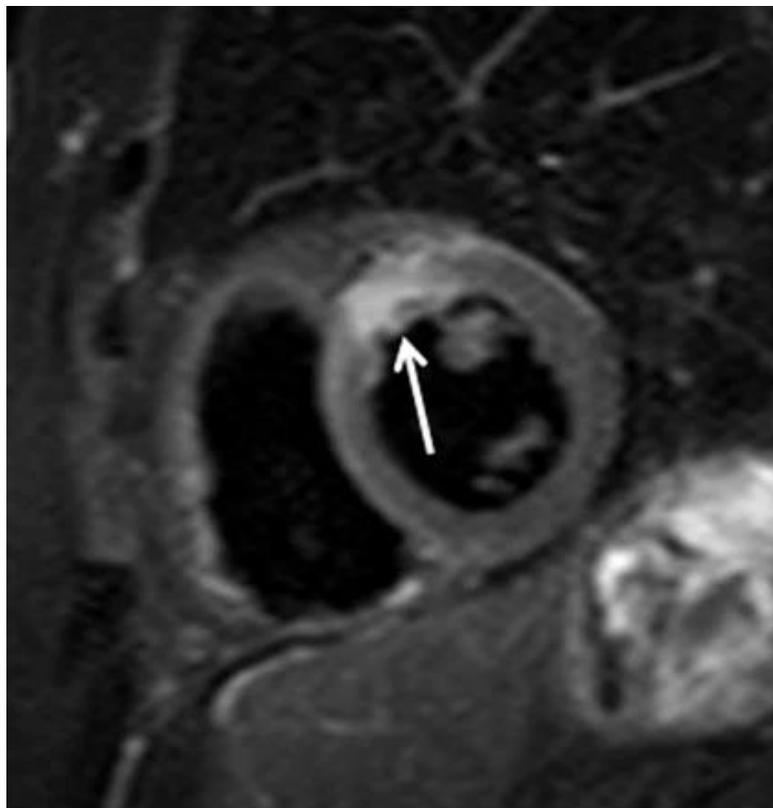
Triple IR SE – Blood And Fat Suppression

- In some cases it is necessary to suppress signal from both blood and fat (e.g. oedema imaging). A third slice selective inversion pulse between the double IR preparation and the image data acquisition can be used to achieve this.
- The inversion time ($T_{I_{fat}}$) of the third inversion pulse is selected so that the longitudinal magnetisation of fat is nulled at the time of data acquisition.
- This techniques is often used with a long TE for T_2 weighting to visualise oedema (as fluid, which has both long T_1 and T_2 , will be bright).





Inversion-Recovery



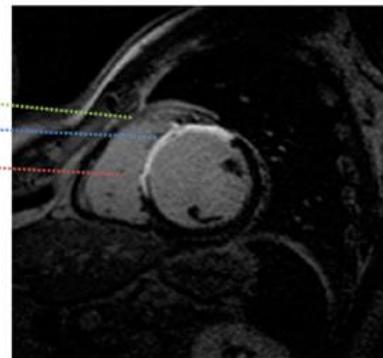
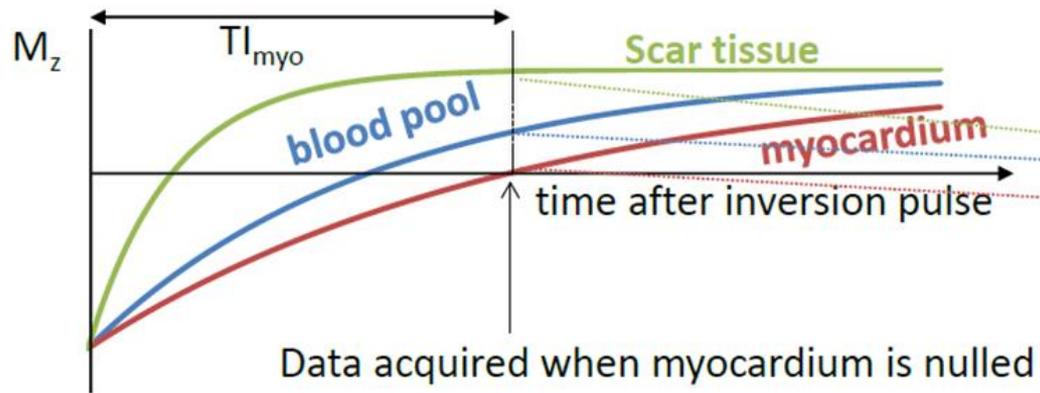


The most commonly used agents are gadolinium based contrast agents, a chelate molecule is required to make the agent non-toxic.

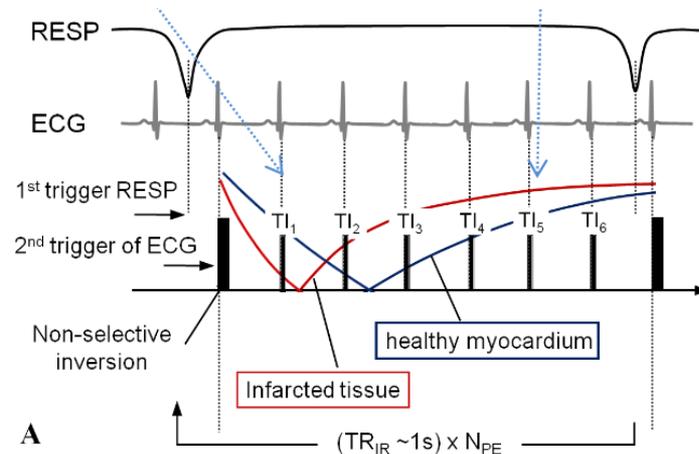
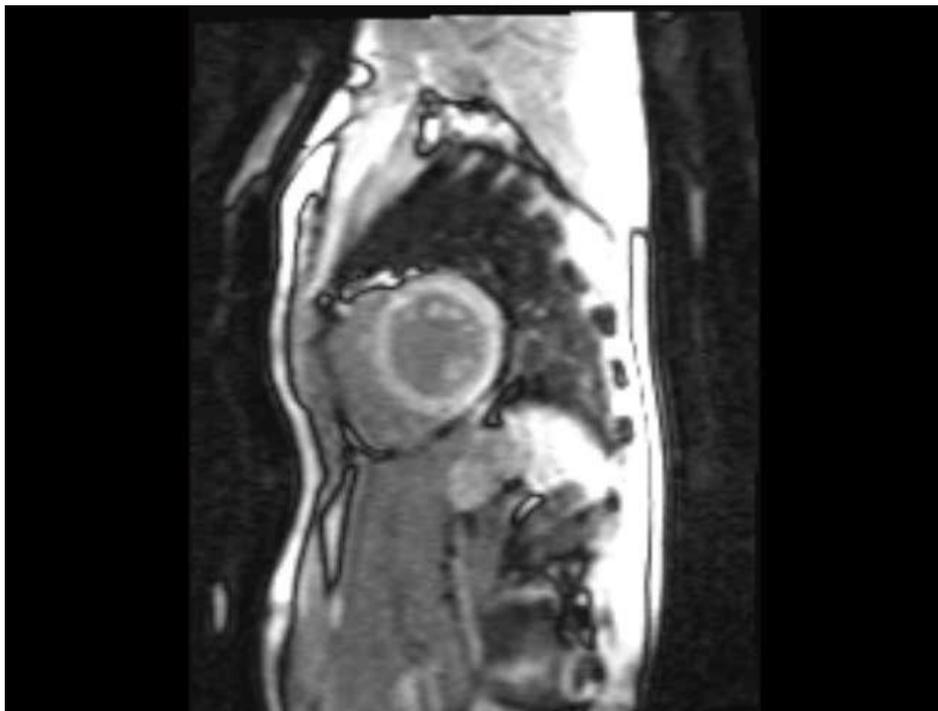
These are injected intravenously and are distributed throughout the body but they are extracellular agents and they leak out of capillaries into interstitial spaces, but do not cross intact cell membranes.

The main effect of GBCAs is to shorten the T1 of the surrounding tissue.

Inversion-Recovery



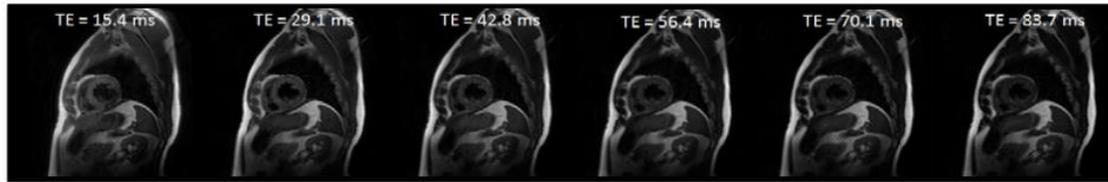
Signal from viable myocardium is nulled in the final image



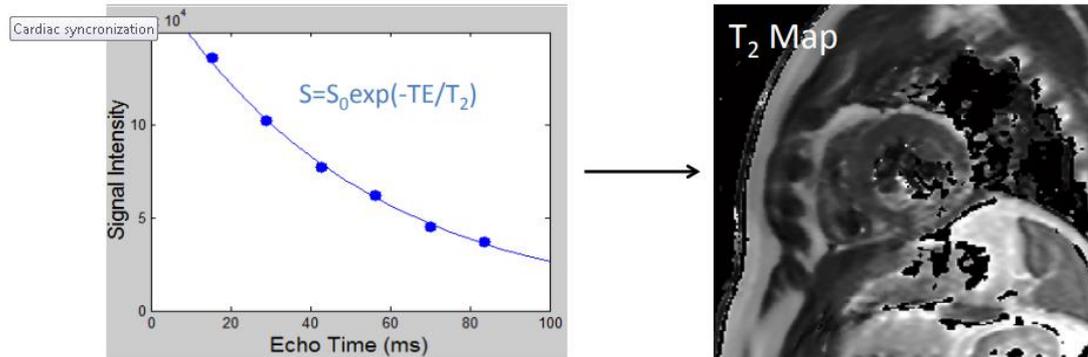
Multiple images are acquired at the same cardiac phase in subsequent R-R intervals after an inversion pulse. This allows images with differing inversion time (TI) values to be acquired at the same cardiac phase

T₂ Mapping Methodology

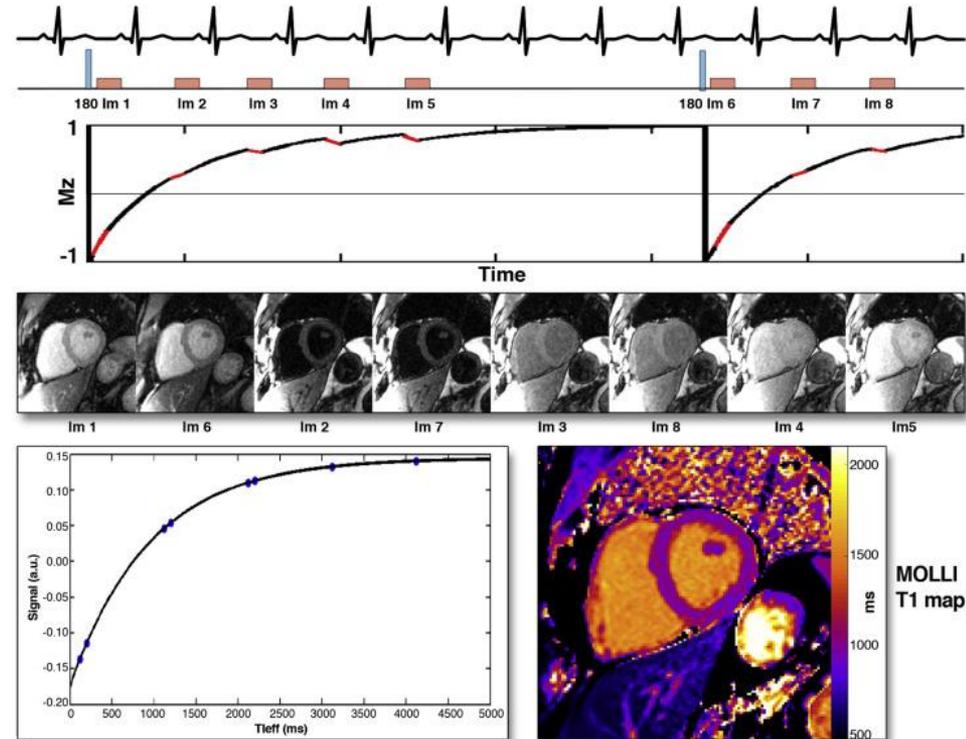
- SE images are acquired with varying TE (and so varying T₂ weighting)



- An exponential decay model is fitted to the acquired signal intensities for each voxel.

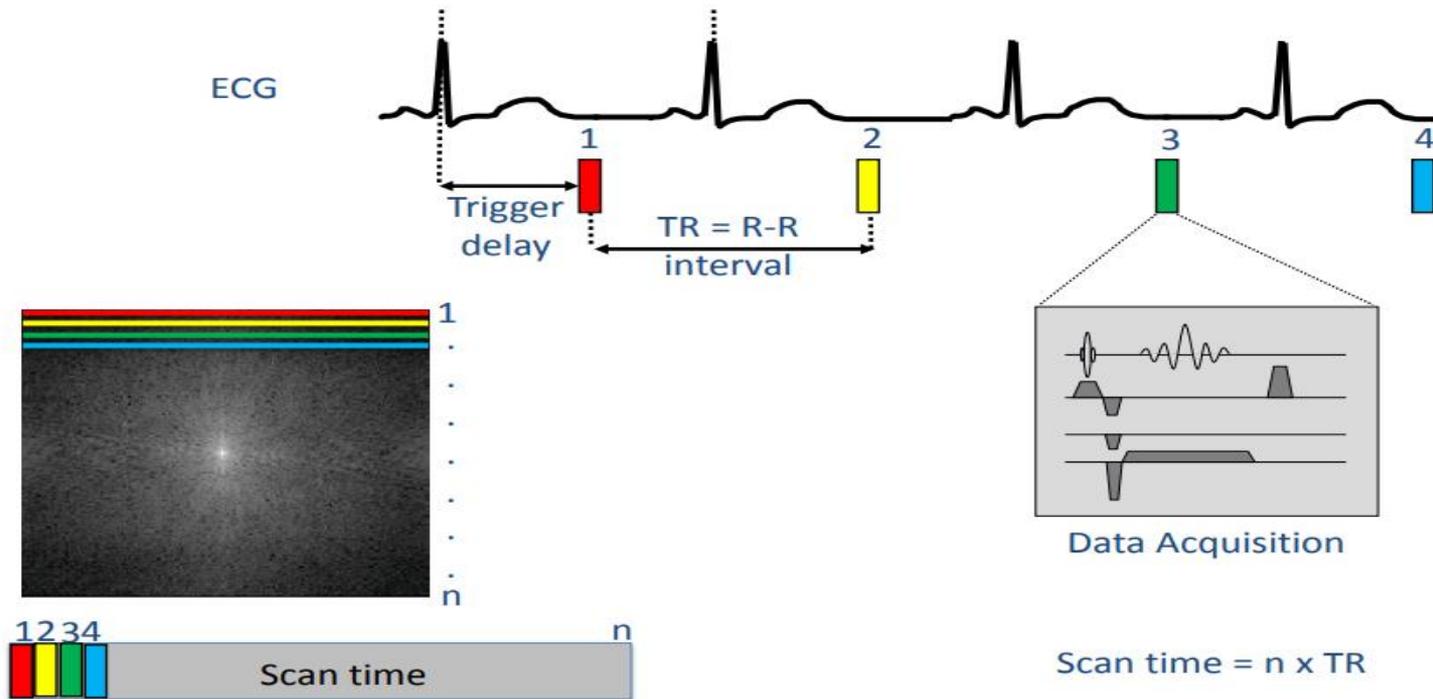


The general principle for T1 mapping is to acquire multiple images with different T1 weightings and to fit the signal intensities of the images to the equation for T1 relaxation



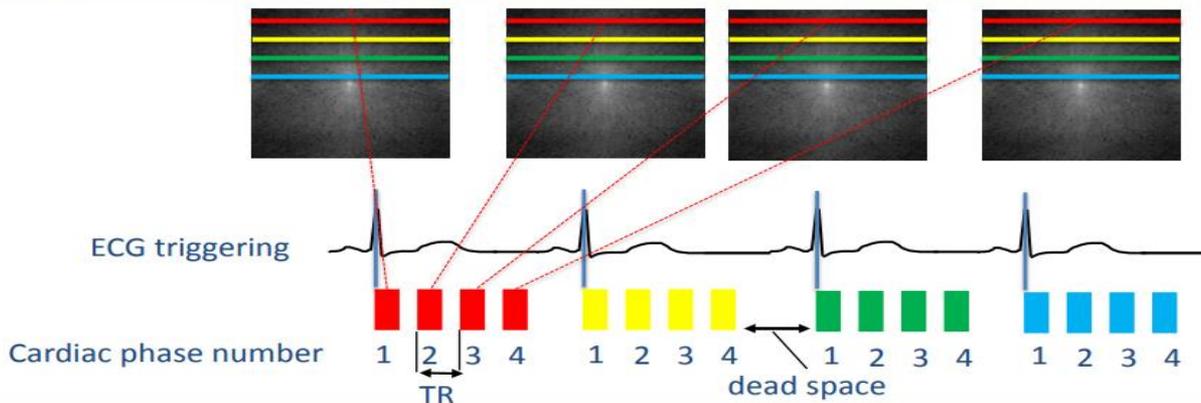
Cardiac synchronization

Conventional ECG Triggered Pulse Sequence

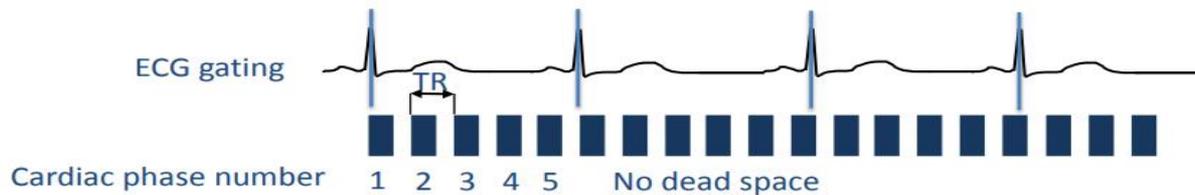


Cardiac synchronization

Cine Imaging With Prospective Triggering



Retrospective Gating

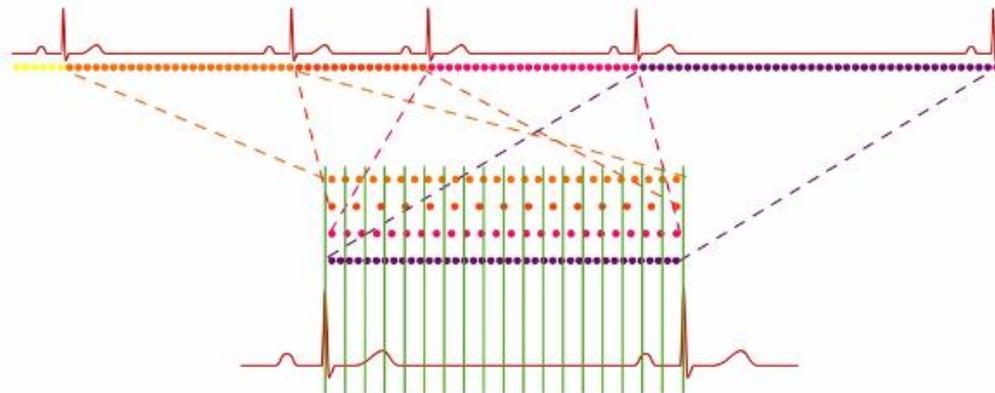


Cardiac synchronization

Prospective ECG gating



Retrospective ECG gating





- Basic Physics

- The creation of MR signal

- The acquisition of the signal

- Image location

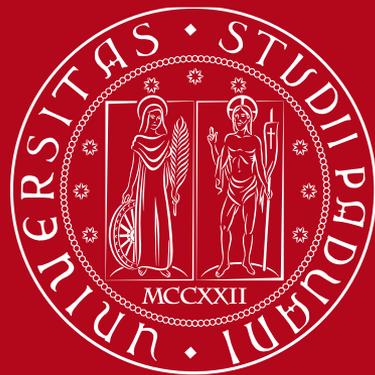
- Image contrast

- Sequences

- Cardiac Syncr



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- Cardiovascular magnetic resonance physics for clinicians: part I. Journal of Cardiovascular Magnetic Resonance volume 12, Article number: 71 (2010)
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