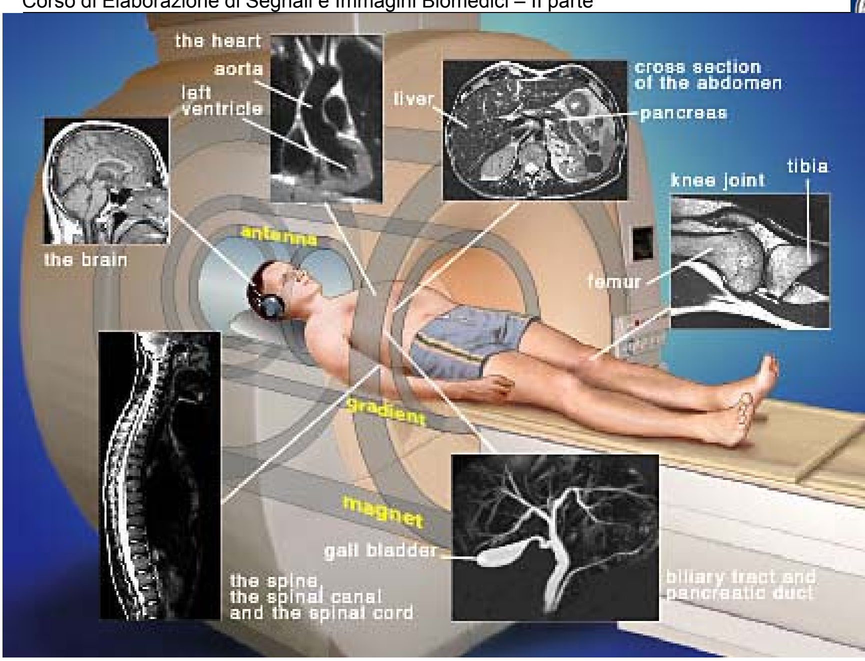
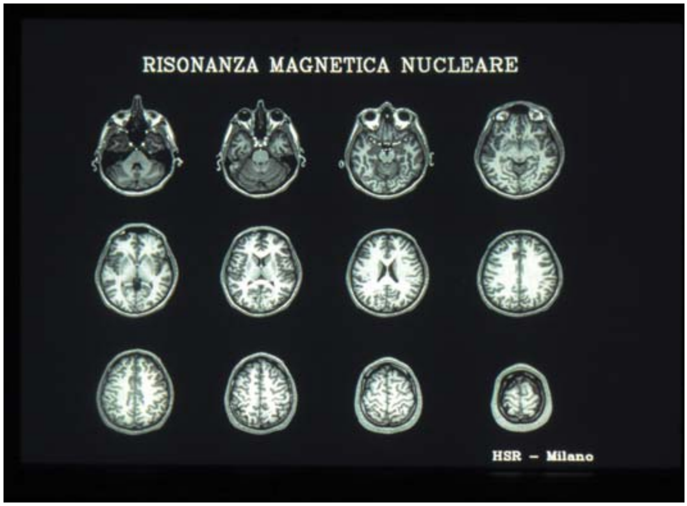
Nuclear Magnetic Resonance Imaging (NMRI)



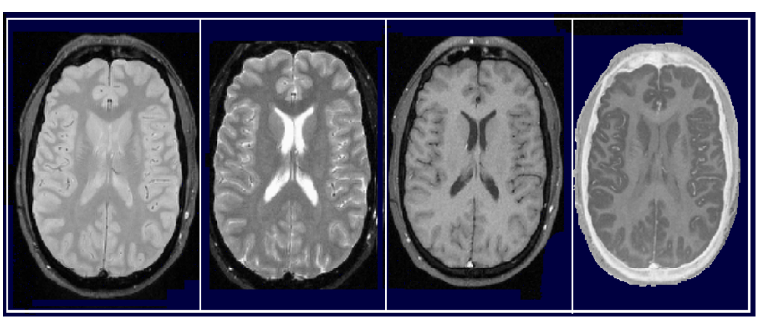
Esempio di MRI

- tomografie assiali MRI encefaliche (cortesia H. S. Raffaele)



Esempio di MRI

- stesso piano tomografico assiale encefalico con diverse metodiche di contrasto MRI (cortesia H. S. Raffaele)



Clinical applications of MRI

Brain MRI:

- It allows to optimally discriminate grey matter (cortex and ganglia), white matter (nervous fibers) and cerebrospinal liquid. Pathological tissues (e.g. multiple sclerosis plaques, brain tumors) can be well distinguished from healthy tissues.
- Whatever section of the brain can be visualized with high resolution. (usually axial and sagittal sections are considered).
- Brain MRI is used as a diagnostic tool and to plan stereotactic surgery.

Spinal MRI:

- It allows to discriminate nerve roots, spinal discs, disc herniation, and spinal cord. The sagittal section allows to visualize the whole vertebral column morphology.

Clinical applications of MRI

Orthopedic MRI: Cartilage and tendons are well contrasted from bones; MRI can substitute explorative arthroscopy.

Abdomen MRI: thanks to the optimal contrast between soft tissues and between soft tissues and liquids and thanks to the high resolution MRI is used to study bile and pancreatic ducts.

Cardiac MRI with gating ECG: high resolution study of cardiac structures and coronaries on an average cardiac cycle.

Oncological MRI: e.g. liver, breast.

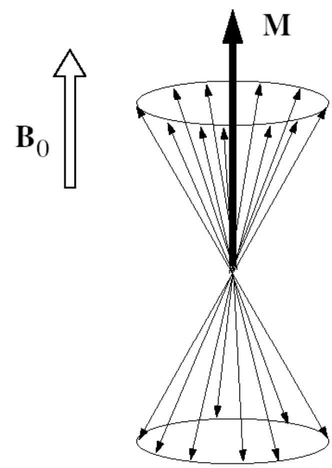
Angio-MRI: blood vessels can be delineated: 1) using Gadolinium based contrast agents; 2) exploiting artifacts produced by moving structures. MRI allows to study vessel anatomy and to extract emodinamic parameters.

Magnetic resonance

• Interaction between **radiofrequency pulses** (10–100 MHz) and biological tissues positioned inside an external **magnetic field B_0**: energy absorption and emission.

• Base of MR:

- – some atoms are provided with a **magnetic dipole;**
- – if positioned inside a magnetic field, the magnetic dipole has a **precession movement** around the field axis.

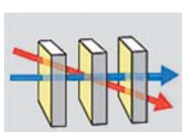


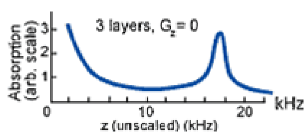
Nuclear magnetic resonance phenomenon

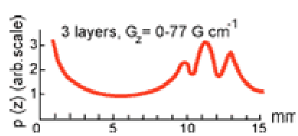
- The discovery is due to Bloch and Purcell (1946, Nobel in 1952).
- It is the resonance of ^1H nuclei (protons) and of other nuclei characterized by an odd number of protons and/or neutrons when exposed to an RF pulse proportional to the external magnetic field intensity.
- Differently from other investigating techniques used in medicine, ionizing radiations are not required. Only magnetic field and RF pulses are used, which can go through tissues without inducing damages.

Magnetic resonance imaging

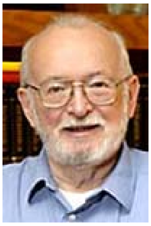
- 1973: Lauterbur understood that the magnetic resonance phenomenon could be employed for medical image generation through the use of **magnetic field gradients**.
- 1980: MRI starts to be used in diagnostic field.
- 2003: Nobel prize to Lauterbur and Mansfield for their MRI studies.
- MRI success and diffusion is mainly due to:
  - Ability to discriminate **soft tissues**;
  - Use of **non ionizing electromagnetic radiation**;
  - Possibility to obtain **images of whatever section** of the object under evaluation;
  - Possibility to obtain **volume information**.
- MR images are obtained from protons in the body (mainly water protons).
- Adopted magnetic field: from 0.5 to 4T (1 Tesla = 10^4 Gauss); more than 20'000 times stronger than the earth's magnetic field (0.25-0.35 Gauss).
- Adopted RF: from 21 to 63 MHz.

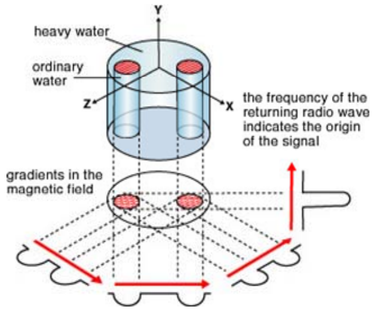






Peter Mansfield discovered that the use of gradients in the magnetic field gave signals that rapidly and effectively could be analysed and transformed to an image. This was an essential step in order to obtain MR images. Mansfield also showed how extremely rapid imaging could be achieved by very fast gradient variations (so called echo-planar scanning). This approach became possible in clinical practice a decade later.





co-ordination of the curves with back-projection calculations results in a transaxial image

Paul Lauterbur discovered that two-dimensional images could be produced by introduction of gradients in the magnetic field. In 1973, he described how addition of gradient magnets to the main magnet made it possible to visualize a cross section of tubes with ordinary water surrounded by heavy water. No other imaging method can differentiate between ordinary and heavy water.

Tesla

- Il **tesla** (simbolo **T**) è una unità di misura derivata del sistema internazionale (SI)
- utilizzata per esprimere la densità del flusso magnetico o l'induzione magnetica (campo magnetico).
- Alla Conference General des Poids et Mesures (CGPM) tenutasi a Parigi nel 1960, venne dato il nome Tesla all'unità di misura, in onore dell'inventore ed ingegnere elettronico Serbo-Americano Nikola Tesla che in vita diede molti importanti contributi nel campo dell'elettromagnetismo.

- ***Nikola Tesla** (cirillico serbo: Никола Тесла) (Smiljan, nell'allora Dalmazia ungherese e odierna Croazia, 10 luglio 1856 - New York, 7 gennaio 1943), fu un fisico, inventore e ingegnere elettrico e meccanico serbo emigrato negli Stati Uniti. **Tesla** è considerato uno dei più importanti inventori della storia. È conosciuto soprattutto per il suo r rivoluzionario lavoro e i suoi numerosi contributi nel campo dell'elettromagnetismo tra la fine dell'Ottocento e gli inizi del Novecento.*

Magnetic resonance – physical principles

• A detailed description of MR requires to use a **quantum mechanical** formalism. However, an easier description can be obtained in terms of **classical mechanics**.

• Every atomic nucleus is associated with a quantum mechanical quantity, called **spin number I** describing properties of the nucleus

• The spin number can assume only certain values (0,1/2, 1, 3/2,…).

• Nuclei having an odd number of protons and/or neutrons have a non zero spin number.

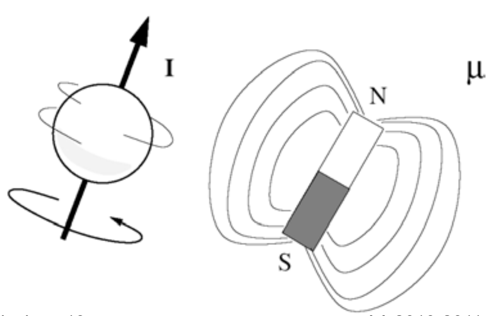
• Nuclei with $I \neq 0$ possess an **angular moment vector I** called **spin**.

Magnetic resonance – physical principles

- In classical mechanics, spin **I** can be represented with a rotation of the nucleus around an internal axis.
- A rotating charge produces a **magnetic moment** $\mu = \gamma \mathbf{I}$ where γ is the **gyromagnetic ratio** of the nucleus. The modulus is:

$$\mu = \gamma \frac{h}{2\pi} \sqrt{I(I+1)}$$

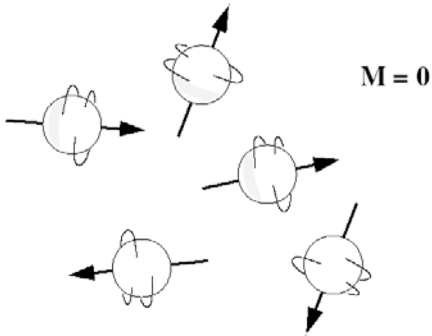
where h=Planck constant



Nuclei can be considered as **magnetic dipoles**.

Magnetic resonance – physical principles

In absence of external magnetic field, dipoles are randomly oriented and therefore the **global magnetization** is **M=**$\sum \mu_i$**=0**.



Magnetic resonance – physical principles

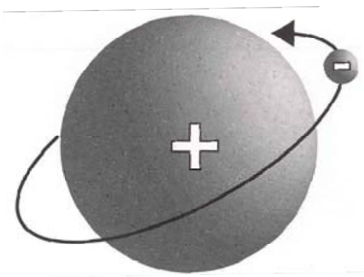
• When a nucleus with spin number I is positioned in an **external static magnetic field B$_0$**, the magnetic moment μ can assume 2I+1 possible orientations corresponding to different energetic levels (Zeeman effect).

• In MR imaging, only nuclei with I**=1/2** and therefore characterized by **only 2 energetic levels** are considered.

Nucleo	$\gamma/2\pi$ (MHz/T)	I(numero di spin)
^{1}H	42.58	1/2
^{2}H	6.53	1
^{13}C	10.71	1/2
^{14}N	3.08	1
^{15}N	4.31	1/2
^{16}O	-	0
^{19}F	40.05	1/2
^{23}Na	11.26	3/2
^{31}P	17.23	1/2
^{39}K	1.99	3/2

Magnetic resonance – physical principles

• **Hydrogen ^{1}H**, thanks to its abundance in biological tissues and thanks to its physical properties which facilitate the detection of MR signals, is the nucleus normally used for MR imaging and will be exclusively considered in the following.



A hydrogen atom consists of a large proton, with positive charge (+), and a smaller electron, with negative charge (−).

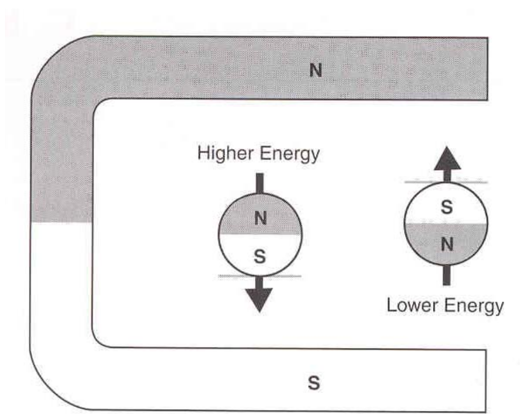


Figure 2-1
Protons can adopt only two states (high and low energy) without any intermediate orientations. Magnetic orientation is therefore "quantized." The preferred low energy state here is indicated with an *arrow* pointing upward, in the direction of the aligning force.

Magnetic resonance – physical principles

- The quantum magnetic numbers corresponding to the $2I+1$ energetic levels are $m_I = -I, \dots, I$, with step=1

  $= (-1/2, 1/2$ per $I=1/2$; $-1, 0, 1$ per $I=1$; etc.)

- The corresponding energy levels are:

  $E = -\mu \cdot \mathbf{B_0} = -\mu_z B_0 = -\gamma (h/2\pi) m_I B_0$ where h=Planck constant.

- For spin with $I=1/2$:

  - <u>parallel</u> to $\mathbf{B_0}$, minimum energy: $E_\uparrow = -1/2\, \gamma\, (h/2\pi)\, B_0$

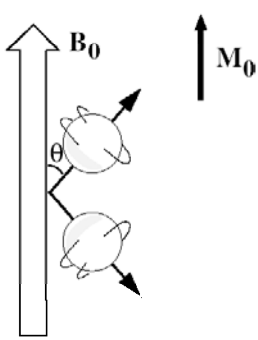
  - <u>anti-parallel</u> to $\mathbf{B_0}$, maximum energy: $E_\downarrow = +1/2\, \gamma\, (h/2\pi)\, B_0$

- Energy difference

  $\Delta E = \gamma (h/2\pi) B_0 = h f_{Larmor}$ (energy of a photon $h\nu$ with $\nu = f_{Larmor}$)

- <u>Larmor frequency</u> $f_{Larmor} = \gamma B_0 / 2\pi$

Magnetic resonance – physical principles



- Magnetic moments corresponding to ^1H take a pointing up (parallel to B_0) or a pointing down (antiparallel to B_0) orientation.
- The **number of parallel** magnetic moments is **larger** than the number of antiparallel magnetic moments .
- The magnetic moment orientation inside the plane orthogonal to $\mathbf{B_0}$ (transverse plane) remains random.
- The result is a **global magnetization** $\mathbf{M_0}$ along $\mathbf{B_0}$ (longitudinal magnetization).

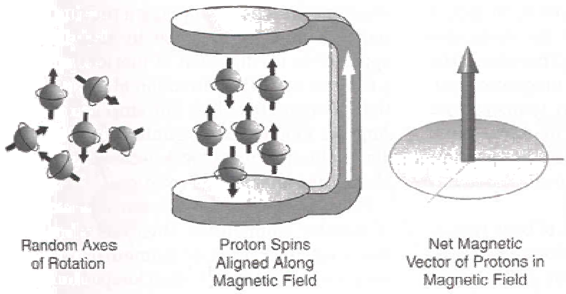
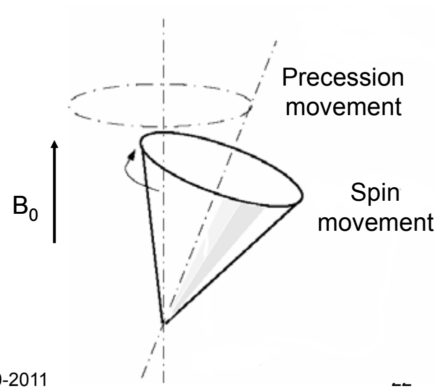


Figure 2-2 Proton rotational axis in the absence of a magnetic field (*left*) is random. In the presence of a magnetic field (*middle*), there is a net excess of protons aligned with the direction of the main magnetic field. This produces longitudinal magnetization (*right*).

Magnetic resonance – physical principles

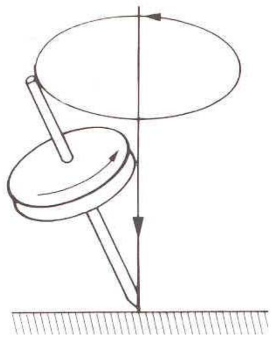
- The external magnetic field $\mathbf{B_0}$, trying to align the magnetic moment μ along the direction of $\mathbf{B_0}$, produces a torque $\mathbf{C}=\mu \times \mathbf{B_0}$
- Rate of change of angular moment vector: $\frac{d\mathbf{I}}{dt} = \boldsymbol{\mu} \times \mathbf{B}_0$
- Motion equation: $\frac{d\boldsymbol{\mu}}{dt} = \gamma\, \boldsymbol{\mu} \times \mathbf{B}_0$

- The result is that the magnetic moment μ precesses around $\mathbf{B_0}$ keeping a constant angle.



Magnetic resonance – physical principles

The same condition subsists when a spinning gyroscope precesses under the effect of gravity.



Magnetic resonance – physical principles

- The precession frequency is proportional to the external magnetic field intensity:

$$\mathbf{f = 1/2\pi \cdot \gamma \cdot B_0}$$ Larmor frequency

- For ^1H, if B_0=1T:

f=42.57 MHz (radiofrequency)

- Precession angular velocity:

$$\Omega_0 = \gamma \cdot B_0$$

The gyromagnetic ratio γ is a characteristic of the nucleus and represents the ratio between the magnetic moment and the angular moment.

Magnetization measurement

• Only nuclei behaving like magnetic dipoles can have an MR response.

• The objective is not to measure the magnetic behavior of single nuclei, but the macroscopic magnetization inside a volume element

$$\mathbf{M} = \sum_i \boldsymbol{\mu}_i$$

• When the system is positioned inside an external static magnetic field $\mathbf{B_0}$, it appears a macroscopic magnetization $\mathbf{M_0}$ oriented like $\mathbf{B_0}$ and with a modulus proportional to the number of magnetic dipole (excited nuclei precessing with their own Larmor frequency).

• A macroscopic magnetization along the direction of $\mathbf{B_0}$ can't be measured. In order to measure the macroscopic magnetization, it must be transferred inside the plane orthogonal to $\mathbf{B_0}$ (transversal plane).

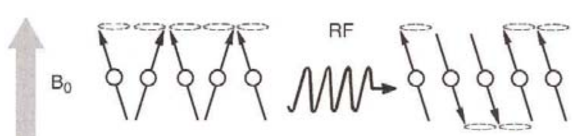
• To do this, a rotating external magnetic field $\mathbf{B_1}$ (Radiofrequency) is added to $\mathbf{B_0}$. $\mathbf{B_1}$ rotates inside the transversal plane with an angular velocity Ω_{B1}. Nuclei with Larmor angular velocity = Ω_{B1} achieve a condition of resonance (from this comes the name of nuclear magnetic resonance).

Effects of the rotating field B1 (RF)

Microscopic level

• The energy required by a dipole to move from the parallel (minimum energy) to the antiparallel (maximum energy) state is $\Delta E = h\, f_{Larmor}$.

• Therefore, the effect of an RF pulse at the Larmor frequency is to gradually change the orientation of dipoles.



• The dipoles don't stop precessing around the direction of $\mathbf{B_0}$ with Larmor frequency!!

Effects of the rotating field B1 (RF)

Macroscopic level

1. Fixed coordinate system

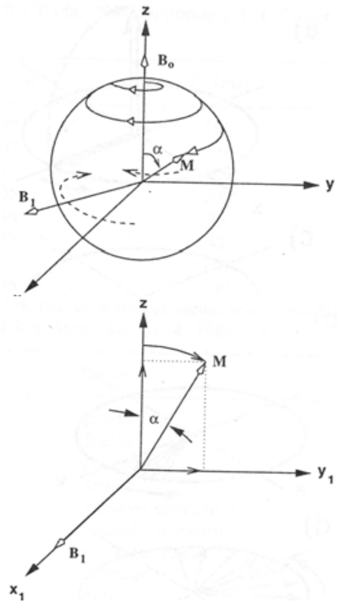
   The global magnetization vector **M** describes a spiral trajectory moving on the surface of a sphere with radius $\mathbf{M_0}$.

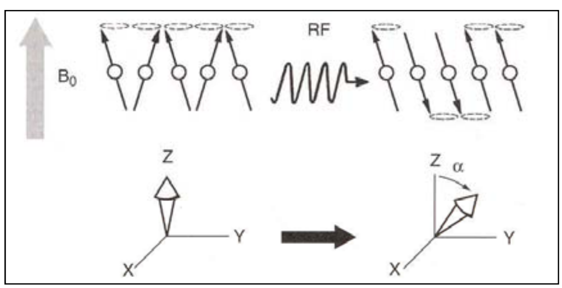
2. Coordinate system firmly rotating with $\mathbf{B_1}$

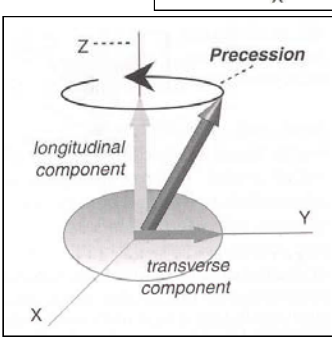
   The global magnetization vector **M** rotates of an angle α around the direction of $\mathbf{B_1}$.

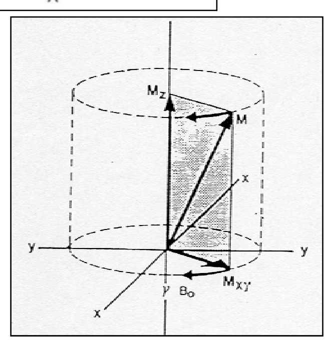
$$\alpha = \Omega\delta t = \gamma B_1 \delta t$$

   A transversal component of magnetization appears!!









Effects of the rotating field B1 (RF)

Changing RF pulses amplitude B_1 and/or duration δt it is possible to make the vector M assuming every orientation.

- 90° pulse: $M_z=0$;
  $M_{xy}=M_0$

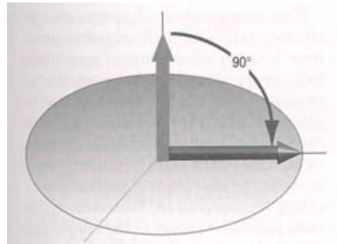
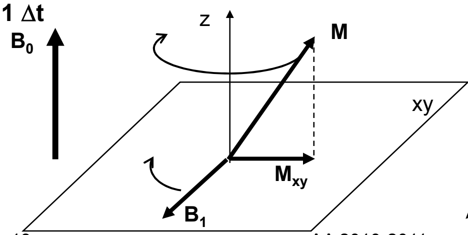


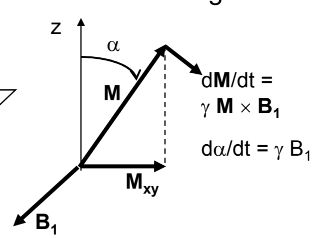
Figure 2-7
A 90° excitation pulse rotates longitudinal magnetization into the transverse plane. In this simplified diagram, net magnetization is represented as vectors (*arrows*).

- 180° pulse: $M_z=-M_0$
  $M_{xy}=0$

Effects of the rotating field B1 (RF)

- The field B1 rotates inside the xy transversal plane at Larmor frequency γB_0 in the direction of the precessing spins. (RF pulse deviates **M** of an angle α, *flip angle,* that increases till the RF pulse is on).
- In the rotating reference frame, B1 produces a second torque $\mu \times$ **B1**. Effects:
  - spins start precessing around B1 with frequency $\gamma B_1 << \gamma B_0$;
  - d**M**/dt = γ **M** $\times$ **B1.** The global magnetization **M** rotates towards the xy plane.
- After an RF pulse applied for a time interval Δt the rotation angle is $\alpha = \gamma$ **B1** Δt





ECCITAZIONE A RADIOFREQUENZA

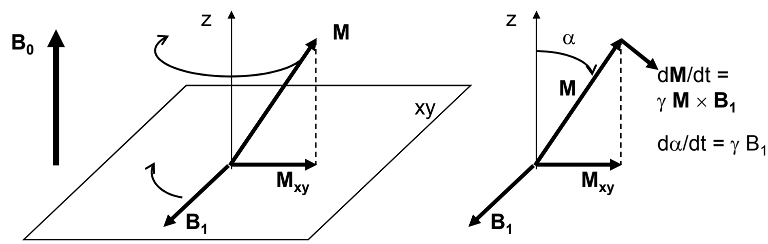
MEDIANTE IMPULSO RF = f_{Larmor}

D'ora in poi facciamo riferimento alla magnetizzazione di volume netta **M**

effetto del contingente di spin-up in eccesso che viene fatto deviare da una RF **in fase in tutto il volume**.

Si consideri che, e.g., per B = 1.5 T, $f_{RF} = f_{Larmor}$ = 63 MHz, λ_{RF} = 5 m

Le RF impiegate sono molto più lunghe delle sezioni di corpo attraversate e quindi in fase in tutto il volume. Inoltre, le antenne impiegate sono antenne di campo vicino (near-field, non antenne di trasmissione far-field) progettate per mantenere la stessa fase nel volume esplorato.



Magnetization measurement

After the application of an RF pulse with flip angle α, the transversal magnetization in the fixed reference frame is:

$$M_{xy}(t) = M_0 \sin\alpha \exp\left[j\left(\Omega_{Larmor} t + \Phi\right)\right] \Rightarrow$$

M_{xy} rotates inside the xy plane at the Larmor frequency

The rotating macroscopic magnetization vector generates an RF signal whose intensity depends on M_0 (proportional to dipole concentration) and on the applied RF flip angle.

$\Downarrow$

<u>Magnetic resonance signal which can be measured!!</u>

MRI –principi generali (1)

- **MRI:** 3D, contrasto tessuti molli, alta risoluzione spaziale (~1mm), non ionizzante
- **Ris temporale**: più lenta di CT e US, ↑ sensibilità movimenti paziente
- **Costo**: elevato. 1,5-T whole body → \$1.5 milioni
- **Principali usi**: brain disease, spinal disorders, angiography, cardiac function, muscoloskeletal damage.

1. Segnale MRI generato da **protoni nel corpo** (acqua e lipidi).
2. Paziente dentro un forte magnete che produce un **campo magnetico statico B_0** > 10.000 volte quello terrestre
3. **Ogni protone** è una particella carica con momento angolare e può essere considerata agire come un **piccolo magnete**.
4. I protoni si allineano in **2 configurazioni** con il loro campo magn. interno **parallelo o anti-parallelo** alla direzione del campo magnetico statico "grande" B_0. Leggermente di piu' nello stato parallelo.
5. **Precessione dei protoni intorno a B_0** come fa un giroscopio influenzato dalla gravità.

MRI –principi generali (2)

6. La frequenza di precessione (e quindi di risonanza) è proporzionale all'intensità del campo magnetico B_0. $f_{LARMOR} = \gamma/2\pi \cdot B_0$
7. L'applicazione di **un debole campo di RF, B_1** provoca una precessione coerente dei protoni; → si rileva la somma di tutti i protoni che precessano come una **tensione indotta in una bobina rivelatore** opportunamente regolata
8. L'informazione spaziale è codificata in immagine usando **gradienti di campo magnetico.** Questi impongono variazioni lineari in tutte le 3 dim del campo magnetico all'interno del paziente.
9. Come risultato di queste variazioni si ha che la **freq di precessione dei protoni dipende linearmente dalla loro posizione nello spazio**.
10. Si misurano Freq e Fase della magnetizzazione dovuta alla precessione con una bobina RF; il segnale analogico è digitalizzato.
11. **Inverse Fourier Transform 2D** converte il segnale nello spazio per produrre un'immagine
12. Variando i parametri di acquisizione si contrastano in modo diverso i tessuti molli

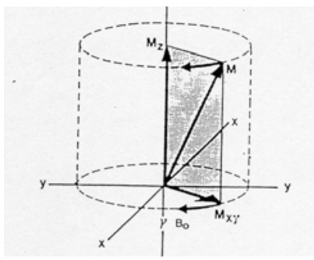
Free Induction Decay (FID)

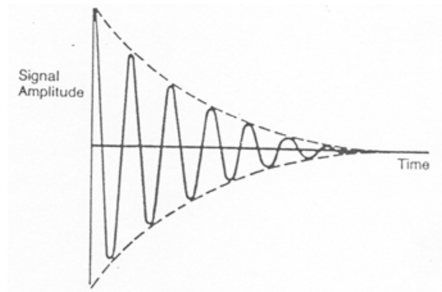
• Once terminated the RF excitation, the system comes spontaneously back to the equilibrium condition (in presence of B_0).

• The longitudinal component Mz of the macroscopic magnetization tends to recover the equilibrium value M_0: longitudinal relaxation

• The transversal component Mxy decays to zero: transversal relaxation.

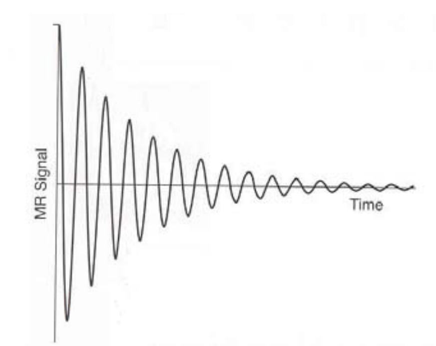
• The signal which can be measured with a receiving coil is the RF generated by Mxy: ***Free Induction Decay (FID).***





Free Induction Decay (FID)

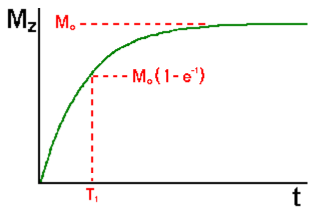
- FID = MR signal detectable after the RF excitation of a magnetic dipole system positioned inside a static magnetic field
- FID's properties:
  - Larmor frequency
  - Amplitude proportional to the relative density of excited nuclei inside the volume under evaluation
  - Decreases exponentially with a time constant depending on the molecular structure containing the excited atoms (relaxation time)

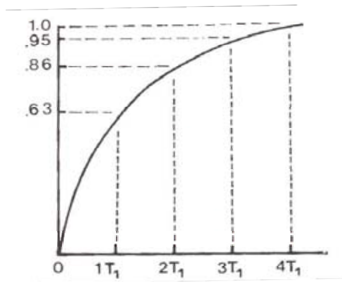


Longitudinal relaxation

Mz tends to come back to the equilibrium value M_0 thanks to the interactions (energy transfers) between spins and surrounding lattice: spin-lattice relaxation with time constant T1.

$$M_z = M_0 \cdot (1-e^{-t/T1}),\ (\alpha=90°)$$





T1=spin-lattice relaxation time constant ≈ 0.05-3 sec (^{1}H)

Longitudinal relaxation

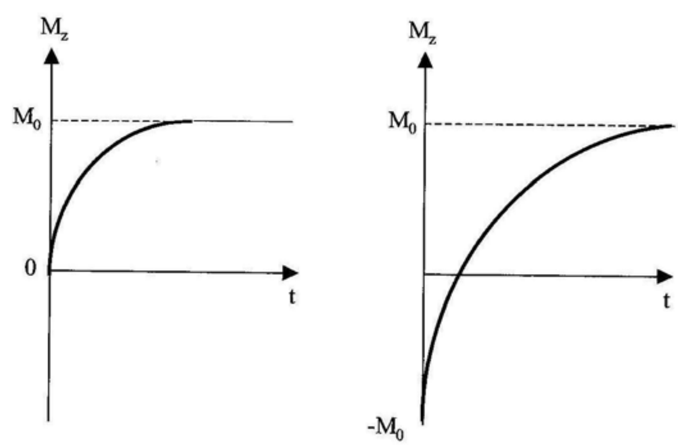
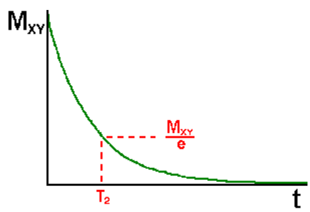


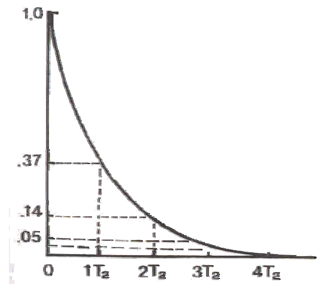
FIGURE 4.8. *Plots of M_z versus time after (left) a 90° pulse and (right) a 180° pulse.*

Transversal relaxation

• Spin-spin interactions generate a loss of phase coherence between precessing protons (different protons precesses with different velocities).
• The macroscopic consequence is a reduction of the magnetization transversal component M_{xy} with time constant T_2.

$$M_{xy} = M_0 \cdot e^{-t/T2}, \quad (\alpha=90°)$$





T_2=spin-spin relaxation time constant ≈ 0.04-2 sec ([1]H)

Transversal relaxation

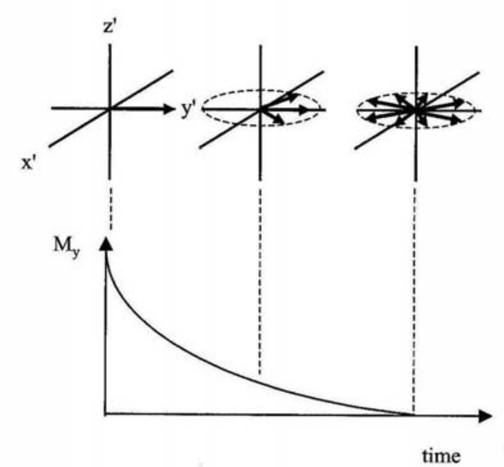


FIGURE 4.9. *(Top) After a 90° pulse, the individual magnetic moments precess at different frequencies because they experience slightly different magnetic fields. (Bottom) The M_y component of magnetization decreases over time, and when the individual vectors are randomly distributed in the transverse plane, there is no net magnetic moment, and no signal is detected.*

Relaxation

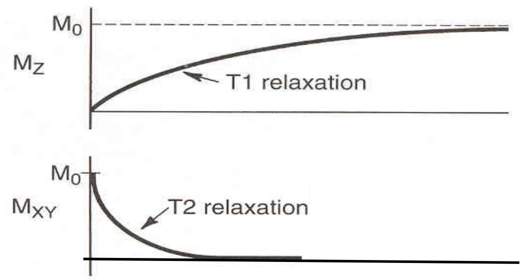


Figure 6-4 After a 90° radiofrequency pulse, M_Z relaxes to M_0. M_{XY} relaxes to zero more quickly because T2 relaxation times are considerably shorter than T1 relaxation times.

Bloch equation

It describes the behavior of the magnetization vector **M** in a fixed reference frame with axis z parallel to $\mathbf{B_0}$.

$$\frac{d\mathbf{M}}{dt} = \gamma \mathbf{M} \times \mathbf{B} - \mathbf{R}\left[\mathbf{M} - \mathbf{M}_0\right]$$

where:

- $\mathbf{B}=\mathbf{B}_0+\mathbf{B}_1$
- **R** = relaxation matrix

$$R = \begin{bmatrix} 1/T_2 & 0 & 0 \\ 0 & 1/T_2 & 0 \\ 0 & 0 & 1/T_1 \end{bmatrix}$$

- T_1 and T_2 depend on molecular mobility, molecular weight, temperature.
- T_1 and T_2 relaxations are not independent .
- $T_2<T_1$; every process taking Mz to its own equilibrium value M_0 makes also Mxy decaying; the vice-versa is not valid.
- The realignment of M with B_0 is not a simple rotation!

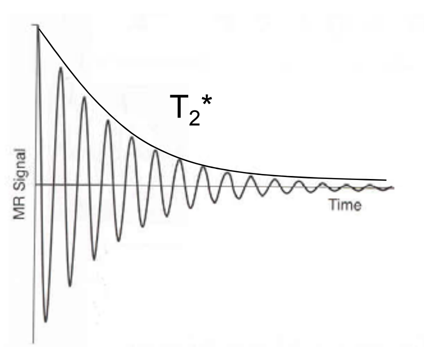
Relaxation time constants for biologic tissues

Tessuto	T_1 (ms)	T_2 (ms)
grasso	192	108
fegato	397	96
pancreas	572	189
tiroide	605	102
muscolo cardiaco	644	75
muscolo scheletrico	629	45
sostanza bianca	687	107
polmone	756	139
milza	760	140
rene	765	124
sostanza grigia	825	110
sangue	893	362
liquor	~ 1500	~ 1500
acqua	~ 3400	~ 3400

Transversal relaxation

Because of inevitable field unhomogeneity, the loss in phase coherence of the transversal magnetization is faster than the one described by T_2.

The time constant describing the real M_{xy} decreasing is $T_2^* < T_2$.



Pulse sequences

• MRI studies objective = to create images where tissues of interest are properly contrasted basing on their **T_1,T_2** or proton density ρ values.

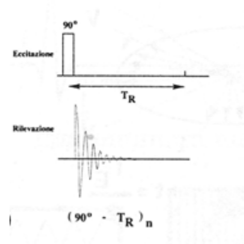
• Pulse sequences allow to empathize the dependence of the measured signal (**FID**) from one of these three parameters.

• All the sequences are characterized by a repetition time **T_R**, which is the time interval between the beginning of a sequence and the successive one.

Saturation recovery

This sequence consist of 90° pulses separated by a repetition time T_R.

• Effect of the first 90° pulse: M_{xy}=M_0 and M_z=0. After the pulse, M_{xy} decays with time constant T_2; M_z increases towards M_0 with time constant T_1>T_2.



• If <u>T_R>T_2 and ≈ T1</u>: $M_{xy}(T_R)$=0, $M_z(T_R)$<M_0.

→A second 90° pulse generates M_{xy}=M_z (T_R) and M_z=0.

A series of 90° pulses allows to study T_1. The amplitude of FIDs after the second one is proportional to:

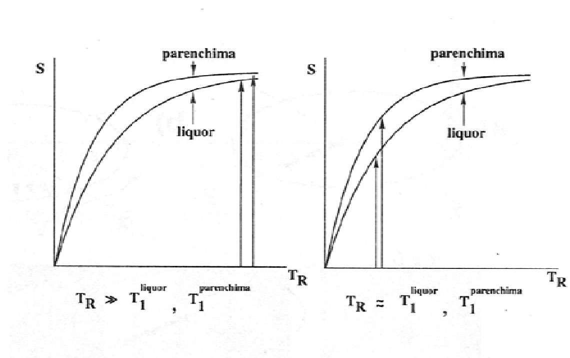
$$\rho\left(1-\exp\left(-\frac{T_R}{T_1}\right)\right)$$

Saturation recovery

• If $T_R > T_2$ and $> T_1$: $M_{xy}(T_R)=0$, $M_z(T_R)=M_0$.

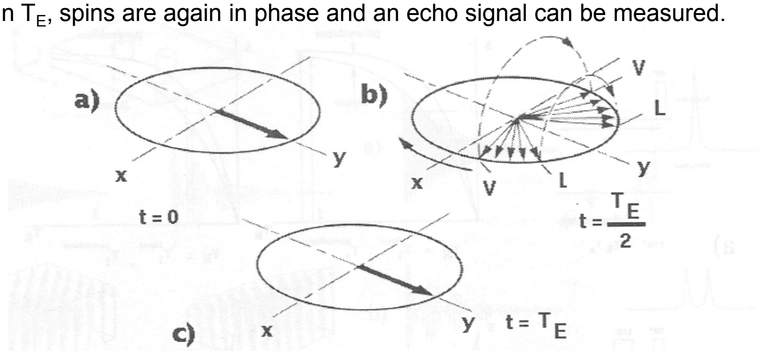
→A second 90° pulse generates $M_{xy}=M_0$ and $M_z=0$.

A series of 90° pulses allows to study the proton density ρ.



Spin echo

This sequence consist of a 90° pulse followed by a 180° pulse after a time interval $T_E/2$.
• 90° pulse effect (t=0): $M_{xy}=M_0$, $M_z=0$.
• After a time interval $T_E/2$ the transversal magnetization M_{xy} is vanished for the spin dephasing effect due to the field inhomogeneity.
• The effect of a 180° pulse in $T_E/2$ is to flip the spins over the other side of the xy plane: gradual recover of spins phase coherence.
• In T_E, spins are again in phase and an echo signal can be measured.



Spin echo

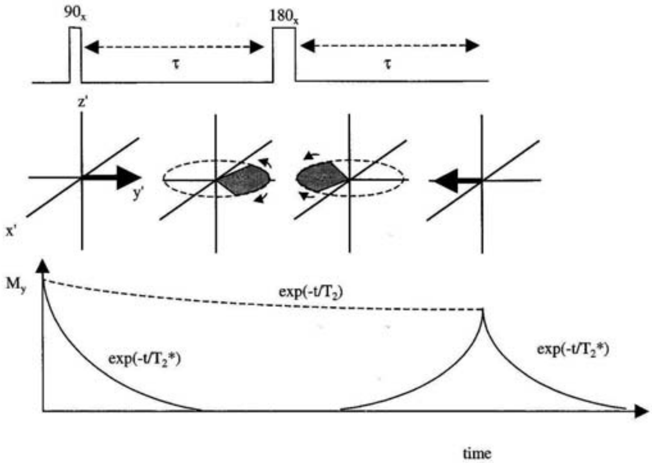
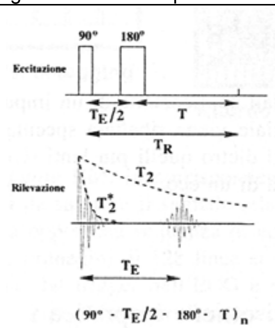


FIGURE 4.10. *A schematic of a spin-echo sequence. The 90° pulse tips the magnetization onto the y axis, where it decays with a time constant T_2^*. The effect of the 180° pulse is to "refocus" the magnetization such that at time τ after the 180° pulse, the individual vectors add constructively, and the signal reaches a peak.*

Spin echo

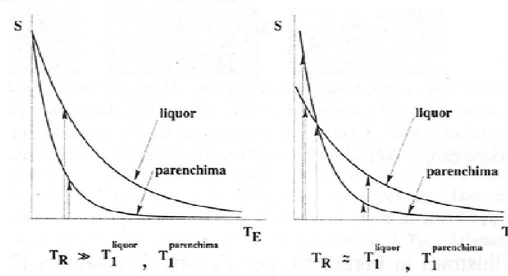
• The echo amplitude is proportional to:

$$\rho \exp\left(-\frac{T_E}{T_2}\right)$$



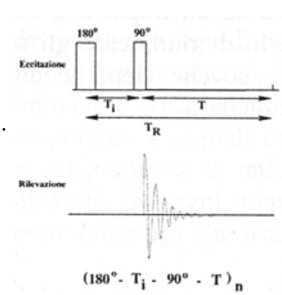
• If a second identical sequence is applied after a repetition time T_R>>T_E, the amplitude of the second echo is proportional to:

$$\rho\left(1-\exp\left(-\frac{T_R}{T_1}\right)\right)\exp\left(-\frac{T_E}{T_2}\right)$$



Inversion recovery

This sequence consist of a 180° pulse followed by a 90° pulse after a time interval T_i.



• Effect of the 180° pulse: $M_{xy}=0$; $M_z=-M_0$. After the pulse, M_z tends to reach the equilibrium value M_0 with time constant T_1.

• After $T_i<T_1$: $M_z(T_i)<M_0$

• Effect of the 90°pulse: $M_z=0$; $M_{xy}=M_z(T_i)$.

• The signal is measured after the 90° pulse and has an amplitude proportional to:

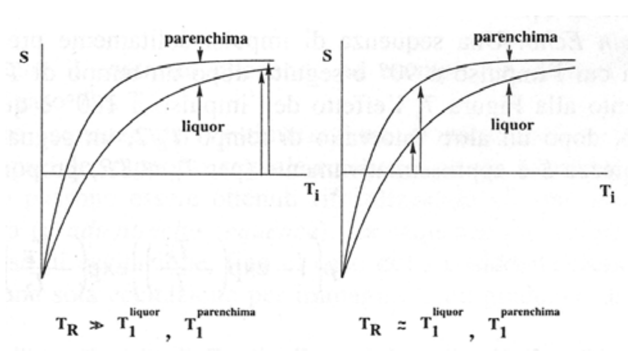
$$\rho\left(1-2\exp\left(-\frac{T_i}{T_1}\right)\right)$$

• If a second identical sequence is applied after a repetition time TR, the amplitude of the second FID is proportional to:

$$\rho\left(1-2\exp\left(-\frac{T_i}{T_1}\right)+\exp\left(-\frac{T_R}{T_1}\right)\right)$$

Inversion recovery

Properly defining the parameters T_i and T_R different contrasts can be obtained.



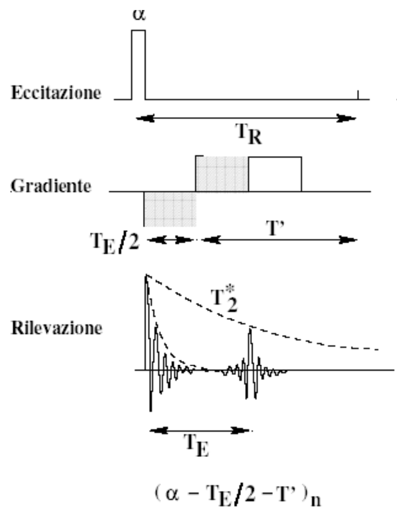
Gradient echo

• Echo signal is generated by using a dephasing/rephasing **magnetic field gradient** after an α pulse.

• The magnetic field gradient induces a linear spatial variation of the magnetic field and therefore a range of resonance frequencies (e.g. $\gamma(B_0\text{-}Gx)$ if a gradient G is applied along x direction): spin dephasing.

• By inverting the gradient, the opposite frequency variation is obtained: spin rephasing.

• By properly choosing α and T_E, different image contrasts can be obtained.



Pulse sequences

• T_1,T_2, ρ and T_{2^*} contrasts all contribute to the image contrast generated by a pulse sequence.

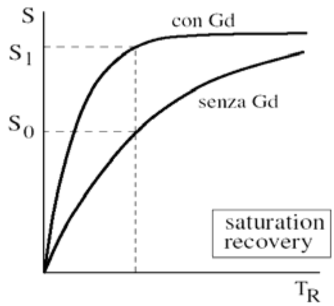
• The sequence parameters must be chosen in order to empathize the contrast of interest.

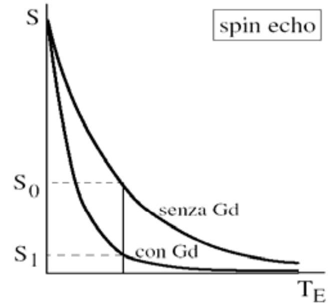
• Advanced sequences exist, able to optimize image characteristics, like spatial resolution, signal to noise ratio, extension of the excited volume, temporal resolution.

• In clinical applications like cardiac studies, imaging of dynamic distribution of contrast agents and brain functional MRI fast sequences are used, like EPI **(Echo Planar Imaging)**.

Contrast agents

- MRI contrast agents are used to increase the contrast between healthy and diseased tissues.
- Two classes of contrast agents exist:
    1. paramagnetic agents (e.g. Gadolinium based), which shorten T1 (and T2) relaxation constants of tissues in which they accumulate;

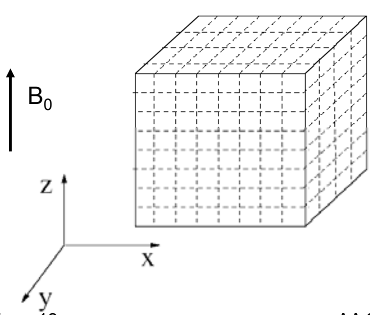


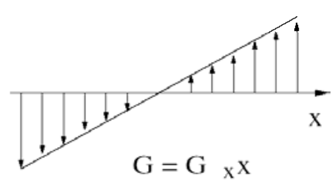


2. Super-paramamagnetic (or ferromagnetic) agents which shorten T2 and T2* values of tissues.

Magnetic resonance imaging

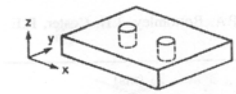
- To create images, the contribute to FID signal of different voxels must be discriminated.
- To discriminate the contribute of different voxels (i.e. spatial positions) the Larmor law is exploited: nuclei are excited with a static magnetic field whose intensity changes with spatial position.
- In order to create a static magnetic field of spatially variant intensity, three magnetic field gradients (G_x, G_y, G_z) are properly added to B_0.
- The intensity of gradients G_x, G_y, G_z is small if compared to the intensity of B_0.



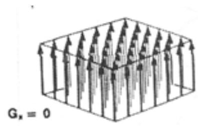


Magnetic resonance imaging: the idea

• Let's consider two test-tubes filled with water and positioned inside a non excitable material.



• If the test-tubes are positioned inside a static magnetic field B_0, they both resonate at the same Larmor frequency γ B_0.



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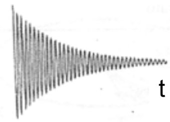
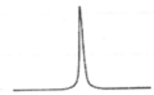


t

FID

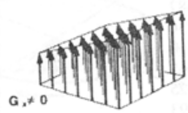
1D Fourier Transform

➡



The two test-tubes can't be discriminated.

• If a magnetic field gradient G_x along the x axis is added to B_0, the Larmor frequencies become $\gamma(B_0+G_x x)$.



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t

FID

1D Fourier Transform

➡

The two test-tubes can be discriminated!

Magnetic resonance imaging: slice selection

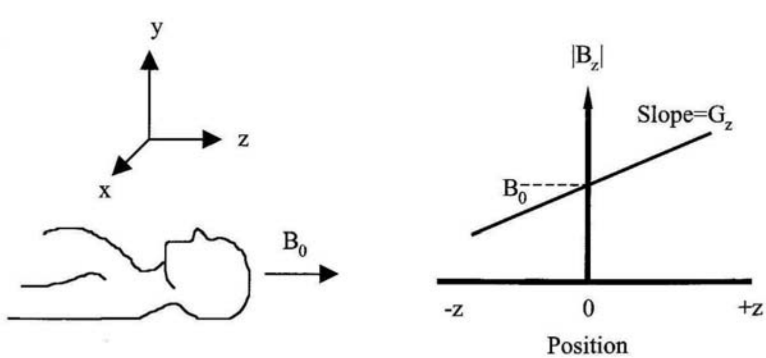
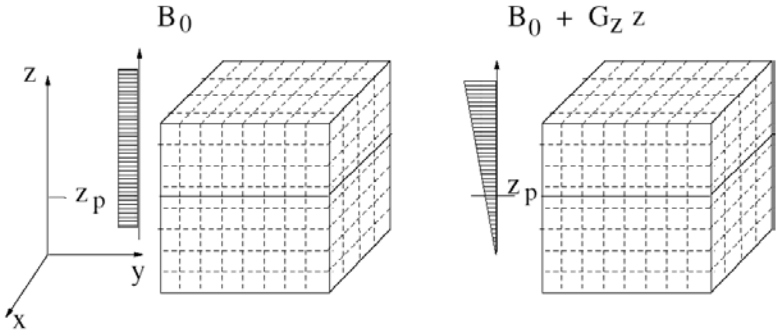


FIGURE 4.15. *(Left) The coordinate system used in the description of all MRI sequences and instrumentation. (Right) A linear magnetic field gradient applied in the z direction produces a linear spatial dependence of the effective magnetic field B_Z.*

Magnetic resonance imaging: slice selection

• A linear magnetic field gradient G_z is applied in z direction.



• All the protons positioned in z_p have Larmor precessional frequency

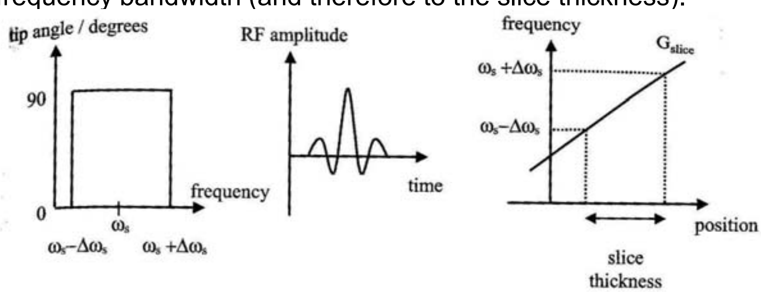
$$\Omega_{zp}=\gamma(B_0+G_z z_p)$$

• An RF pulse with frequency Ω_{zp} excites only nuclei with coordinate $z=z_p$.

Magnetic resonance imaging: slice selection

Shape and thickness of the excited slice depend on G_z and on the RF pulse.

- Ideal RF pulse: pulse with rectangular frequency profile $\omega_s \pm \Delta\omega_s$
  – It excites protons within a rectangular-shaped slice;
  – Slice thickness $\Delta z=2\Delta\omega_s / \gamma 2\pi\, G_z$;
  – The RF pulse has a sinc envelope function: $B_1(t) \propto \mathrm{sinc}(t/\Delta t)$
  – The RF pulse length (e.g. 1-5msec) is inversely proportional to the frequency bandwidth (and therefore to the slice thickness).



- Real RF pulse: e.g. Gaussian with Gaussian frequency profile and slice shape.

Magnetic resonance imaging: slice selection

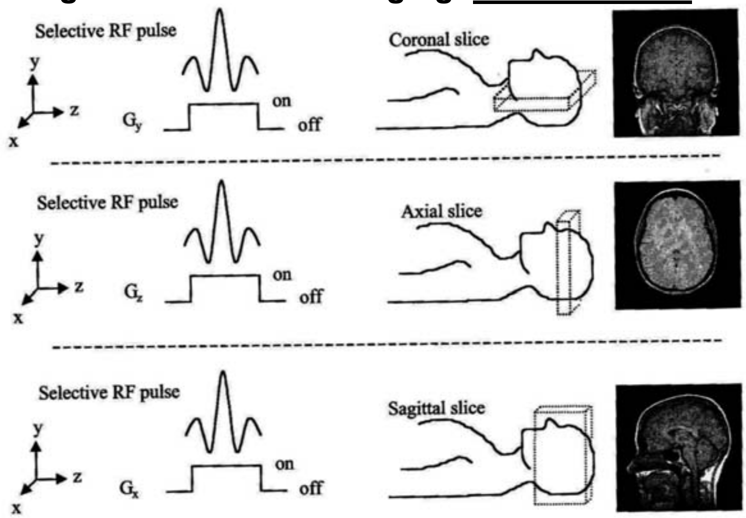


FIGURE 4.16. A schematic of slice selection in MRI. By using a frequency-selective pulse in combination with the y, z, or x gradient, a coronal, axial, or sagittal slice, respectively, can be chosen.

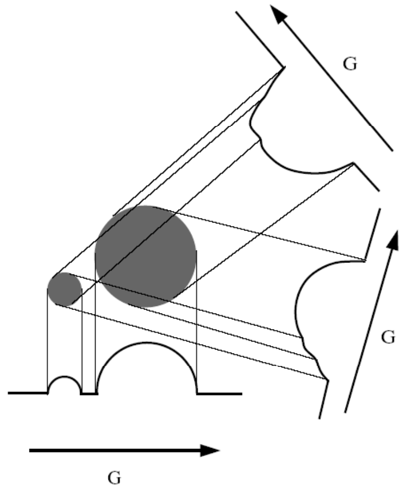
Magnetic resonance imaging: 2D image construction

There are two possible strategies:

1) Generation of 1D projections of the 2D proton distribution (frequency encoding) + 1D Fourier Transform + image reconstruction: strategy adopted in first MRI scanners.

2) Frequency encoding and phase encoding + 2D Fourier Transform of the acquired data.

2D image construction through generation of projections

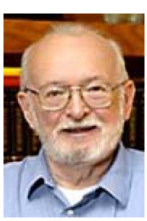


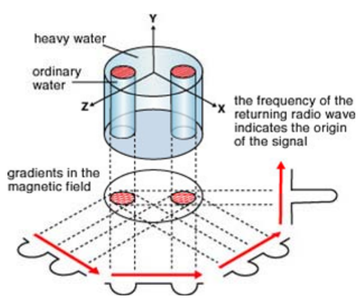
By acting on the gradient components along x and y, a gradient oriented along whatever orientation can be obtained.

$$\Phi = arctg \frac{G_y}{G_x}$$

The 1D Fourier Transform of the acquired FID is a projection of the 2D proton distribution.

2D image construction through generation of projections

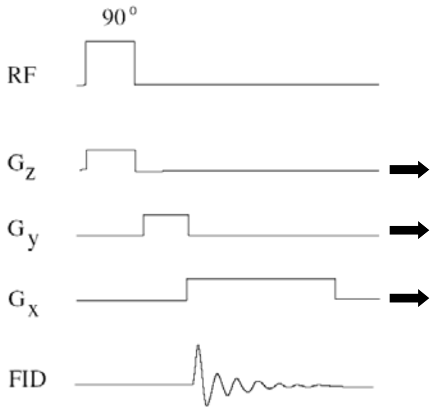




co-ordination of the curves with back-projection calculations results in a transaxial image

Paul Lauterbur discovered that two-dimensional images could be produced by introduction of gradients in the magnetic field. In 1973, he described how addition of gradient magnets to the main magnet made it possible to visualize a cross section of tubes with ordinary water surrounded by heavy water. No other imaging method can differentiate between ordinary and heavy water.

2D image construction: frequency encoding and phase encoding



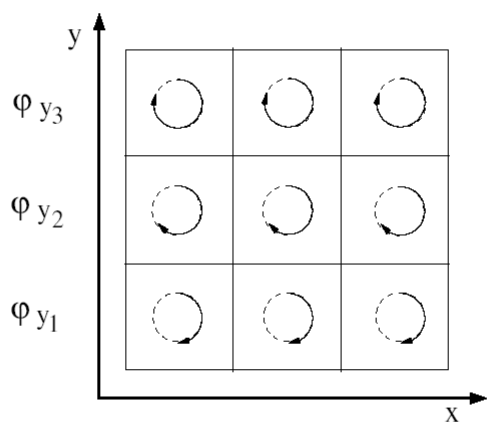
Slice selective gradient, applied during RF stimulation

Phase encoding gradient, applied after RF stimulation and before FID acquisition

Frequency encoding gradient, applied during FID acquisition

Phase encoding

- A magnetic field gradient G_y is applied for a period t_y and switched off before FID acquisition.
- During G_y application the protons precess with frequency γG_y y.
- In $t=t_y$ protons come back precessing with frequency γB_0, but do maintain a phase shift depending from their position along axis y: $\varphi_y = \gamma\, G_y\, y\, t_y$



Frequency encoding

• During FID acquisition, a magnetic field gradient G_x is applied in the x direction.

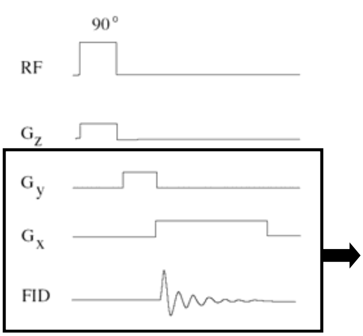
• The protons precess with a frequency dependent from their position along x axis: $\gamma G_x x$.

• The measured FID is:

$$FID(t) \propto \iint \underbrace{I(x,y)\exp(j\gamma G_y y t_y)}_{\text{Phase encoding}} \underbrace{\exp(j\gamma G_x x t)dxdy}_{\text{Frequency encoding}}$$

where I(x,y) is the information of interest, depending from proton density, T1 and T2.

2D image construction: frequency encoding and phase encoding

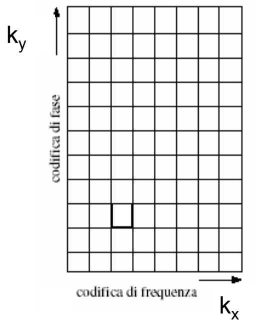


• These operations are repeated N_y times, where N_y is the number of rows of the image to be obtained.

• The intensity of the phase encoding gradient is $n_y G_y$, where $n_y=1, \dots, N_y$.

• The measured FIDs are digitalized in N_x samples, where N_x is the number of columns of the image to be obtained. Sampling interval = t_x.

2D image construction: frequency encoding and phase encoding



If we define:

$$k_y = n_y \gamma G_y t_y$$

$$k_x = n_x \gamma G_x t_x$$

The n_y-th digitalized FID is:

$$FID_{k_y}(k_x)$$

$$= \iint_{x,y} I(x,y)\exp(jk_x x + jk_y y)dxdy$$

The RM image *I(x,y)* can be obtained by applying the 2D Fourier Transform to the matrix of collected data!!

Multislice imaging

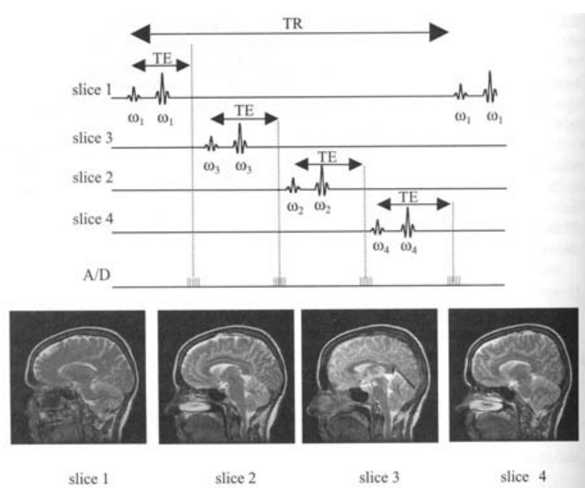
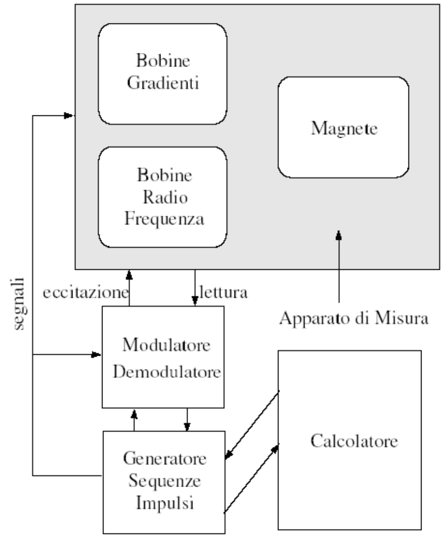


FIGURE 4.31. (Top) A spin-echo multislice imaging sequence. The slices are acquired from different positions by adjusting the frequency offset ω_n of the RF pulses. Only the RF waveforms are shown; the slice, phase phase-, and frequency-encoding gradient waveforms are identical to those in Figure 4.27. In this example, four slices can be acquired in each TR interval. (Bottom) Four sagittal slices acquired using the multislice sequence.

MRI instrumentation



MRI instrumentation

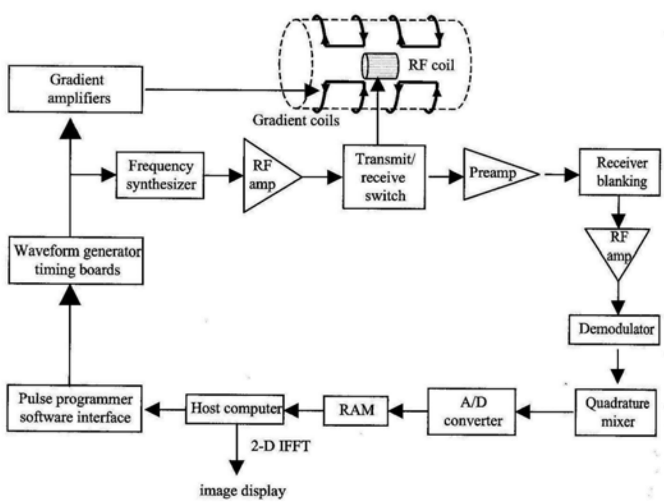


FIGURE 4.21. *A block diagram of the electronic and computer components making up an MRI system.*

MRI instrumentation: the magnet

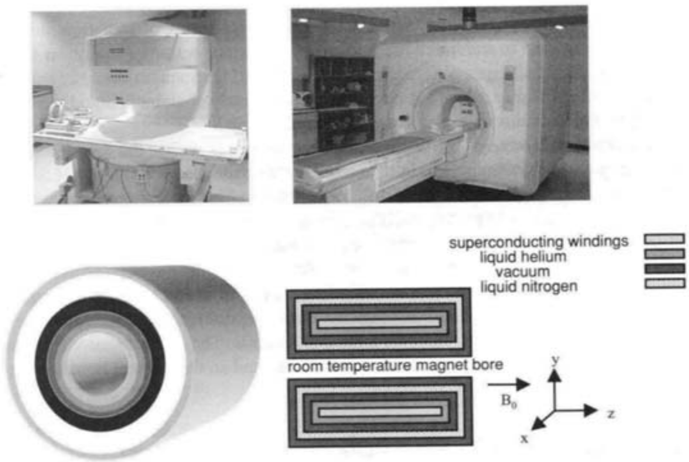


FIGURE 4.22. *(Top left) An open "C-arm" permanent magnet operating at 0.3 T. The magnet has two pole pieces, one above and one below the patient bed, an arrangement that allows easy access to the patient. (Top right) A clinical superconducting MRI magnet operating at a magnetic field strength of 1.5 T. (Bottom) A schematic of the construction of a superconducting magnet.*

MRI instrumentation: magnetic field gradient coils

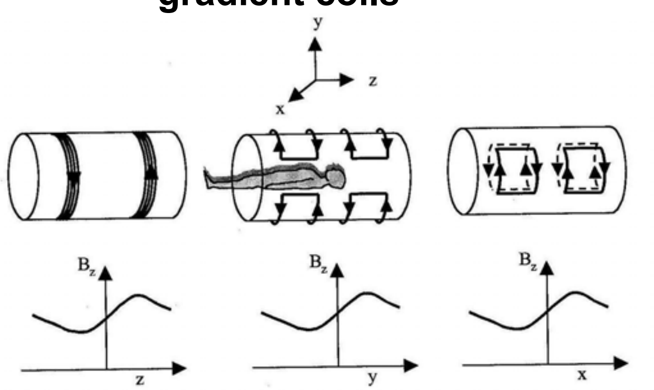


FIGURE 4.23. *The basic design of magnetic field gradient coils used for MRI. The arrows indicate the direction of current flow. (Left) A z-gradient coil, (center) a y-gradient coil, and (right) an x-gradient coil. Each coil consists of multiple turns of wire, which for clarity are only shown for the z-gradient coil. The useable region of the magnet effectively corresponds to the volume over which the gradients are linear.*

MRI instrumentation

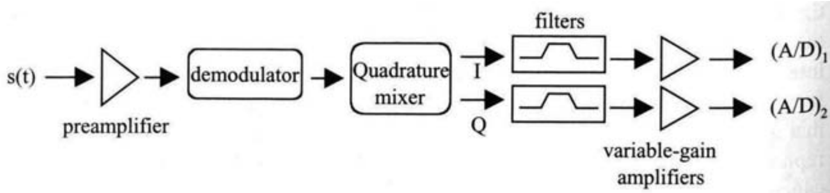


FIGURE 4.11. *A block diagram of the receiver used in NMR and MRI systems. The demodulator reduces the frequency of the signal from the Larmor frequency to an intermediate frequency, typically 10.7 MHz. The quadrature mixer separates the real and the imaginary components of the signal, demodulated to the "baseband" frequency. These components are bandpass-filtered, amplified, and fed into two A/D converters.*

MRI instrumentation

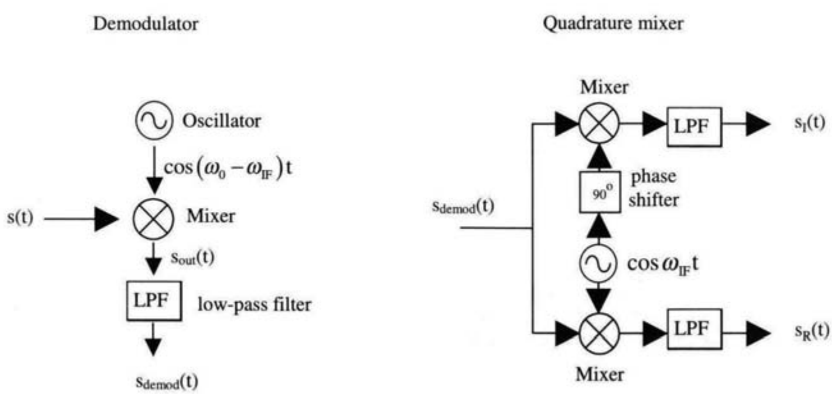


FIGURE 4.12. *(Left) A circuit diagram for a demodulator which takes the signal s(t) at the Larmor frequency and outputs a signal $s_{demod}(t)$ at an intermediate frequency ω_{IF}. (Right) A circuit for a quadrature demodulator which outputs the real and the imaginary components of the signal at baseband frequency.*

MRI instrumentation: RF coils

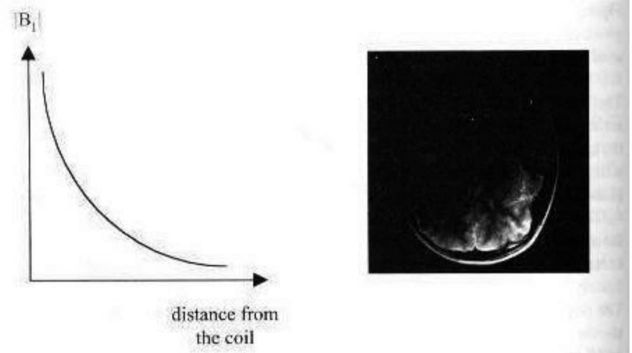


FIGURE 4.26. (Left) A plot of the magnitude of the B_1 field versus distance from the coil for a surface coil. Very high signal can be obtained close to the coil, but the signal from tissues deep in the patient is very low. (Right) An image obtained using a surface coil placed close to the back of the head.

Functional magnetic resonance imaging (fMRI)

- It is based on different magnetic properties of oxygenated hemoglobin (diamagnetic) and deoxygenated hemoglobin (paramagnetic).
- During cerebral activity, both blood flow (oxygenated hemoglobin supply) and oxygen metabolic rate locally increase.
- The oxygen consumption is superior respect to the supply.
- Therefore, during neural activity, the ratio of oxygenated/deoxygenated hemoglobin increases respect to the steady state.
- The MRI signal intensity is sensitive to this ratio: Blood Oxygen Level Dependent (BOLD) Effect.
- The deoxygenated hemoglobin increases the field inhomogeneity. Therefore during neural activation an increase in T2* can be measured.
- fMRI is used to localize brain areas involved in cognitive tasks.

Chemical shift (e.g., fat vs. water) implies very small frequency shifts (ppm) which can be normally overlooked while using gradients which cause shifts 2-3 orders of magnitude larger.

However some artifacts can appear (e.g., fat artifact) which can be attenuated by chemical shift sensitive sequences.

Conversely chemical shift imaging (CSI) or MR spectroscopy (MRS) do emphasize chemical effects.

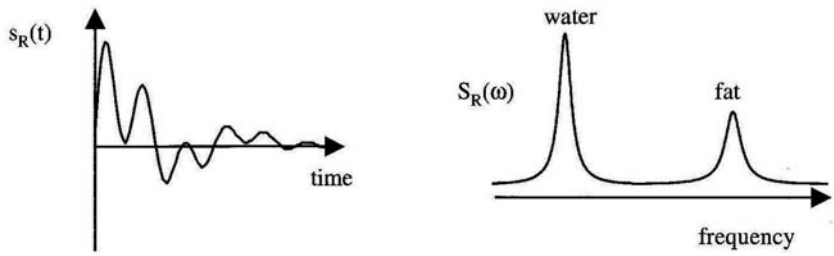


FIGURE 4.14. *(Left) The real component of the time-domain signal corresponding to the proton signals from water and fat. (Right) The real component of the NMR spectrum obtained by Fourier transformation of the time-domain signal.*