الجامعة التقنية الوسطى
كلية التقنيات الصحية والطبية / بغداد
قسم : تقنيات الاشعة المادة: الفيزياء الشعاعية / الرنين المغناطيسي المرحلة: الثالثة

Title: **العنوان:**

Historical Introduction

Name of the instructor: **اسم المحاضر:**

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Target population: **الفئة المستهدفة:**

3rd year طلبة المرحلة الثالثة

قسم تقنيات الاشعة

Posttest: **الاختبار البعدي:**

References: **المصادر:**

Thayalan, K., and Ramamoorthy Ravichandran. *The physics of radiology and imaging*. JP Medical Ltd, 2014.

Historical Introduction

Magnetic resonance imaging (MRI) is an imaging technique used primarily in medical settings to produce high quality images of the inside of the human body. MRI is based on the principles of nuclear magnetic resonance (NMR), a spectroscopic technique used by scientists to obtain microscopic chemical and physical information about molecules. The technique was called magnetic resonance imaging rather than nuclear magnetic resonance imaging (NMRI) because of the negative connotations associated with the word nuclear in the late 1970's. MRI started out as a tomographic imaging technique, that is it produced an image of the NMR signal in a thin slice through the human body. MRI has advanced beyond a tomographic imaging technique to a volume imaging technique. This package presents a comprehensive picture of the basic principles of MRI.

- Before beginning a study of the science of MRI, it will be helpful to reflect on the brief history of MRI. Felix Bloch and Edward Purcell, both of whom were awarded the Nobel Prize in 1952, discovered the magnetic resonance phenomenon independently in 1946. In the period between 1950 and 1970, NMR was developed and used for chemical and physical molecular analysis.
- In 1971 Raymond Damadian showed that the nuclear magnetic relaxation times of tissues and tumors differed, thus motivating scientists to consider magnetic resonance for the detection of disease.
- In 1973 the x-ray-based computerized tomography (CT) was introduced by Hounsfield. This date is important to the MRI timeline because it showed hospitals were willing to spend large amounts of money for medical imaging hardware. Magnetic resonance imaging was first demonstrated on small test tube samples that same year by Paul Lauterbur. He used a back projection technique similar to that used in CT.

- In 1975 Richard Ernst proposed magnetic resonance imaging using phase and frequency encoding, and the Fourier Transform. This technique is the basis of current MRI techniques. A few years later, in 1977, Raymond Damadian demonstrated MRI called field-focusing nuclear magnetic resonance. In this same year, Peter Mansfield developed the echo-planar imaging (EPI) technique. This technique will be developed in later years to produce images at video rates (30 ms / image).
- Edelstein and coworkers demonstrated imaging of the body using Ernst's technique in 1980. A single image could be acquired in approximately five minutes by this technique. By 1986, the imaging time was reduced to about five seconds, without sacrificing too much image quality. The same year people were developing the NMR microscope, which allowed approximately 10 μm resolution on approximately one cm samples. In 1987 echo-planar imaging was used to perform real-time movie imaging of a single cardiac cycle. In this same year Charles Dumoulin was perfecting magnetic resonance angiography (MRA), which allowed imaging of flowing blood without the use of contrast agents.
- In 1991, Richard Ernst was rewarded for his achievements in pulsed Fourier Transform NMR and MRI with the Nobel Prize in Chemistry. In 1992 functional MRI (fMRI) was developed. This technique allows the mapping of the function of the various regions of the human brain. Five years earlier many clinicians thought echo-planar imaging's primary applications were to be in real-time cardiac imaging. The development of fMRI opened up a new application for EPI in mapping the regions of the brain responsible for thought and motor control. In 1994, researchers at the State University of New York at Stony Brook and Princeton University demonstrated the imaging of hyperpolarized 129Xe gas for respiration studies.

- In 2003, Paul C. Lauterbur of the University of Illinois and Sir Peter Mansfield of the University of Nottingham were awarded the Nobel Prize in Medicine for their discoveries concerning magnetic resonance imaging. MRI is clearly a young, but growing science.

Why MRI?

When using x-rays to image the body one doesn't see very much. The image is gray and flat. The overall contrast resolution of an x-ray image is poor. In order to increase the image contrast one can, administer some sort of contrast medium, such as barium or iodine-based contrast media. By manipulating the x-ray parameters kV and mAs one can try to optimize the image contrast further but it will remain sub optimal. With CT scanners one can produce images with a lot more contrast, which helps in detecting lesions in soft tissue.

The principal advantage of MRI is its excellent contrast resolution. With MRI it is possible to detect minute contrast differences in (soft) tissue, even more so than with CT images. By manipulating the MR parameters one can optimize the pulse sequence for certain pathology. Another advantage of MRI is the possibility the make images in every imaginable plane, something, which is quite impossible with x-rays or CT. (With CT it is possible to reconstruct other planes from an axially acquired data set).

However, the spatial resolution of x-ray images is, when using special x-ray film, excellent. This is particularly useful when looking at bone structures. The spatial resolution of MRI compared to that of x-ray is poor. In general, one can use x-ray and CT to visualize bone structures whereas MRI is extremely useful for detecting soft tissue lesions. Before beginning a study of the science of MRI, it will be helpful to reflect on the brief the hardware of MRI.

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Title: **العنوان:**

The Hardware of MRI

Name of the instructor: **اسم المحاضر:**

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1.1 The Hardware of MRI

Scanners of magnetic resonance imaging (MRI) come in many varieties. There is a permanent magnet type, resistive, superconducting, and opening or bore, with or without helium, high field strength or low. The choice of magnet mainly governed by what you intend to do, and the cost. Field magnets offer high quality image better, faster scanning and a wider range of applications, but they are more cost than their counterpart's field is low.

1.2 Magnet Types

The static magnetic field (B_o) in MRI systems can be created by: **Permanent magnets** and **Electromagnets.**

1.2.1 Permanent Magnets

A **permanent magnet** that originates from permanently ferromagnetic materials, which does not lose the magnet field, that remains over time without weakening. Due to weight considerations, **these types of magnets** are usually limited to maximum field strengths of 0.4 T (the unit for magnetic field strength is Tesla: 1 Tesla = 10000 Gauss).

ADVANTAGES	DISADVANTAGES
Low power consumption	Limited field strength (<0.3T)
Low operating cost	Very heavy
Small fringe field	No quench possibility
No cryogen	

Figure 1: Open MRI system "OPER"

Permanent magnets have usually an open design system (see Figure 1) which has ample open space which is more comfortable for the patient. So, the open design accommodates extremely large patients and dramatically reduces anxiety for all patients especially those who have claustrophobic tendencies or have larger body structures.

1.2.2 Electromagnets

There are two categories can be used in MR scanner: Resistive and Superconducting Magnets

1.2.2.1. Resistive Magnets

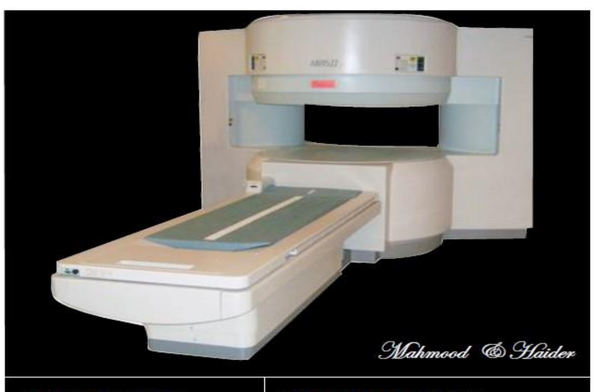
Resistive magnets are made from loops of wire wrapped around a cylinder through which a large electric current is passed. These magnets are very large that utilizes the principles of electromagnetism to generate the magnetic field, like the ones used in scrap yards to pick up cars. They are lower in cost, but need a lot of power to run that means, large current values which runs through loops of wire because of the natural resistance of the wire. Therefore, they produce a lot of heat, which requires significant cooling of the magnet coils. Resistive magnets come in two general categories: iron-core and air-core. Resistive magnets are typically limited to maximum field strengths can be up to 0.6 Tesla. They usually have an open design, which reduces claustrophobia. Figure 2 shows Hitachi's Airis 0.3 Tesla (air-core) system.

1.2.2.2 Superconducting Magnets

Superconducting magnets are Today's most commonly used in MRIs. These superconductors, such as niobium-tin and niobium-titanium are used to make the coil windings for superconducting magnets. The magnetic field is generated by a passing electrical current through coils of wire. The wire is surrounded with a coolant, such as liquid helium, to reduce the electric resistance of the wire. At 4 Kelvin (-269 C^{o}) electric wire loses its resistance (Figure 3). Once a system is energized, it won't lose its magnetic field. Superconductivity allows for systems with very high field strengths up to 12 Tesla. The ones that are most used in clinical environments run at 1.5 Tesla. Most superconducting magnets are bore type magnets. A number of vacuum vessels, which act as temperature shields, surround the core. These shields are necessary to prevent the helium to boil off too quickly. Another advantage of superconducting magnets is the high magnetic field homogeneity

[In 1997 Toshiba introduced the world's first open superconducting magnet. The system uses a special metal alloy, which conducts the low temperature needed for superconductivity. The advantage of this is that the system does not need any helium refills, which dramatically reduces running costs. The open design reduces anxiety and claustrophobia. Figure 9.4 shows Toshiba's OPART 0.35 Tesla system, which combines an open design with the advantages related to superconducting magnets.]

The current trend in magnet design is low field open design versus high field bore design. Obviously, it would be desirable to combine the two, and only time will tell whether this can be done within reasonable manufacturing costs and technical/structural limitations.



ADVANTAGES	DISADVANTAGES
Low capital cost Light weight Can be shut off	High power consumption Limited field strength (<0.2T) Water cooling required Large fringe field

ADVANTAGES	DISADVANTAGES
High field strength High field homogeneity Low power consumption High SNR Fast scanning	High capital costs High cryogen costs Acoustic noise Motion artifacts Technical complexity

Title: العنوان:

SHIMMING

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1.3 Shimming

MRI requires a very high homogeneous static magnetic field. In order to produce high-resolution images, the magnetic field inhomogeneity produced in a high-performance MRI scanner must be maintained to the order of several ppm. After manufacturing, the magnet must be adjusted in some points to produce a more uniform field by making small mechanical and/or electrical adjustments to the overall field. This process is known as **shimming**. Because the magnet itself is not adequately homogeneous, it is necessary to improve or "shim" the homogeneity of the static magnetic field (Bo). A **shim** is a device used to adjust the homogeneity of a magnetic field.

Shimming (or adjustment of the static magnetic field homogeneity) is accomplished by two methods: (1) Passive shimming (2) Active shimming

- **Passive shimming:** The mechanical adjustments, which add small pieces of iron or magnetized materials, are typically called passive shimming. Passive shimming involves pieces of steel with good magnetic qualities. The steel pieces are placed near the permanent or superconducting magnet. They become magnetized and produce their own magnetic field.
- **Active shimming:** The electrical adjustments, which use extra exciting currents, are known as active shimming. Active shimming is performed with coils with adjustable current. Active shimming requires passage of electric current through coils with unique geometric configurations. The shim coils are designed to correct inhomogeneities of specific geometries.

In both cases (active and passive shimming), the additional magnetic fields (produced by coils or steel) add to the overall magnetic field of the superconducting magnet in such a way as to increase the homogeneity of the total field.

1.4 Radio Frequency Coils

Radio Frequency (RF) coils are needed to receive and/or transmit the RF signals used in MRI scanners. RF coils system comprises the set of components for transmitting and receiving the radiofrequency waves involved in exciting the nuclei, selecting slices, applying gradients and in signal acquisition. RF coils are vital component in the performance of the radiofrequency system. They one of the most important components that affects image quality and obtaining clear images of the human body. RF coils for MRI can be categorized into two different categories: volume coils and surface coils.

1.4.1 Volume RF Coils

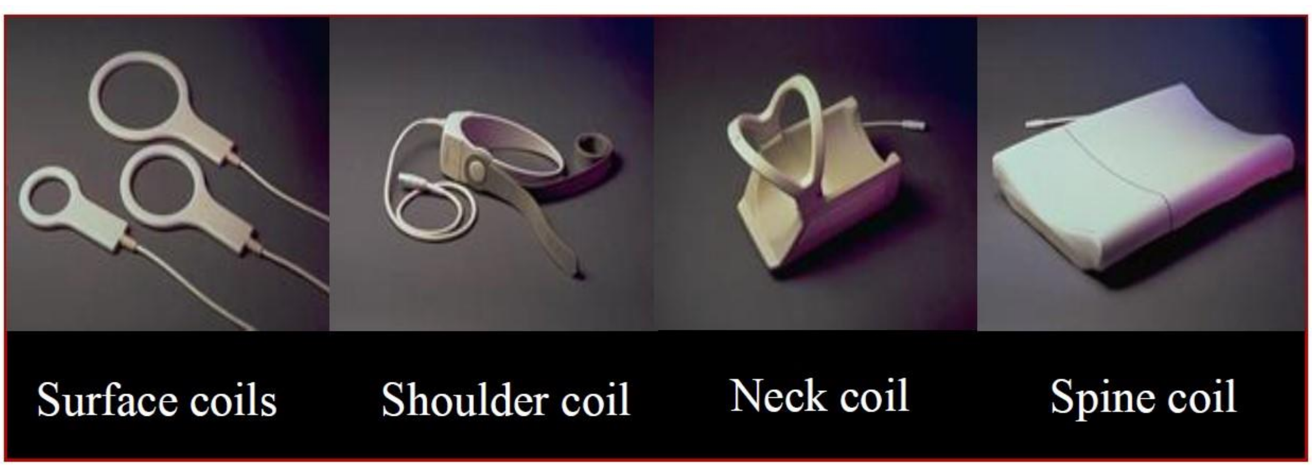
The design of a volume coil is to provide a homogeneous RF field inside the coil which is highly desirable for transmit, but is less ideal when the region of interest is small. The large field of view of volume coils means that by receiving the noise that they receive from the whole body, not just the region of interest. Volume coils need to have the area of examination inside the coil. They can be used for transmit and receive, although sometimes they are used for receive only.

Figure 4: shows two volume coils (a) Head coil (b) Knee coil

Most clinical applications volume coil is built to perform whole-body imaging, and smaller volume coils have been constructed for the head and other extremities. These coils are requiring a great deal of RF power because of their size, so they are often driven in quadrature in order to reduce by two the RF power requirements. Figure 5 shows two volume coils. The head coil is a transmit/receive coil; the knee coil is received only.

1.4.2 Surface Coils

Surface coils have very high RF sensitivity over a small area of interest. As the name already implies, surface coils are placed over or around the surface of the anatomy of interest to the patient directly such as the temporo-mandibular joint, the orbits or the shoulder. The coil consists of single or multi-turn loops of copper wire. They have a high ***Signal to Noise Ratio*** *(SNR)* and allow for very high-resolution imaging because their small field of view and hence they only detect noise from the region of interest. The disadvantage is that they lose signal uniformity very quickly when you move away from the coil. In case of a circular surface coil, the depth penetration is about half its diameter. Surface coils make poor transmit coils because they have poor RF homogeneity, even over their region of interest. Figure 6 shows a few examples of surface coils.

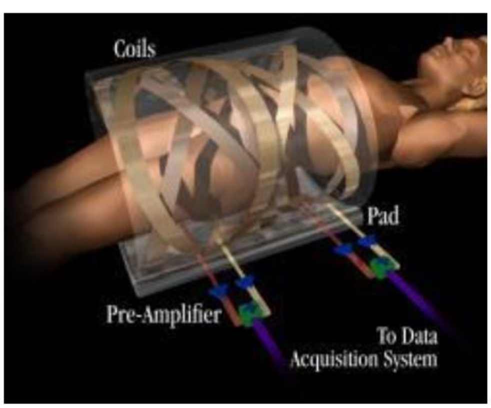


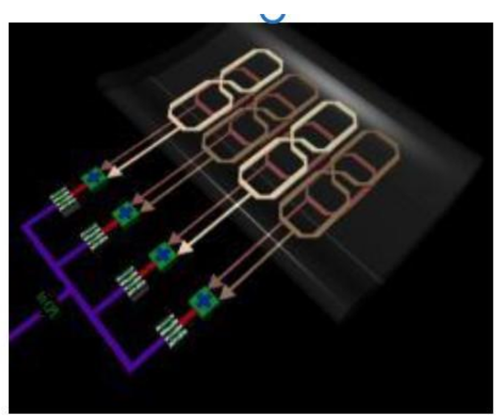
1.4.3 Quadrature Coils

The quadrature coil consists of two coils, which are placed at right angles to one another that mean oriented 90 degrees relative to each other. Therefore, the MRI signals received by each, coil is 90 degrees out of phase with each other. The advantage of this design is that they produce √2 more signal than single loop coils. The quadrature coil operates in the circular polarization circularization mode. The quadrature coil can generate three types of images: Real image, Imaginary image, and Magnitude image. Nowadays, most volume coils are Quadrature coils. The coils shown in (Figure 6) are Quadrature coils.

1.4.4 Phased Array Coils

Phased array coils consist of multiple surface coils with small diameter which are combined (coil elements in phased array) to record the signal simultaneously and independently, so a greater level can be explored. Surface coils have the highest signal-to-noise ratio (SNR) than that delivered by one large diameter but have a limited sensitive area. By combining 4 or 6 surface coils it is possible to create a coil with a large sensitive area. (Figure 7) shows the design of two phased array coils. The QD Body Array coil is a volume coil, while the Spine Array coil is a surface coil. Phased Array coils produce in average √2 more signal than Quadrature coils. Today most MRI systems come with Quadrature and phased array coils.





1.5 Other Hardware

There is more hardware needed to make an MRI system work. A very important part is the Radio Frequency (RF) chain, which produces the RF signal transmitted into the patient, and receives the RF signal from the patient (see figure 8). Actually, the receive coil is a part of the RF chain.

Faraday shield

The frequency range used in MRI is the same as used for radio transmissions. That's why MRI scanners are placed in a **Faraday** cage to prevent radio waves to enter the scanner room, which may cause artifacts on the MRI image. Someone once said: "MRI is like watching television with a radio". To function properly, an MRI scanner needs to sit in a specialized room or chamber shielded against Radio Frequency (RF) interference. Without such protection the very weak RF signals that emanate from the patient when scanned would be overwhelmed. Also, to stop the radio frequencies produced by the scanner from interfering with equipment outside the cage. Furthermore, one needs a processor to process the received signal, as well as to control the complex business of scanning.

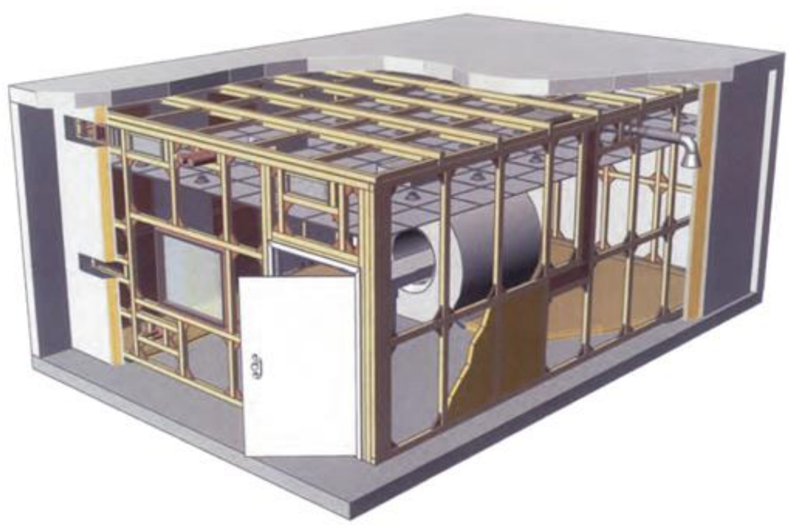


Figure 8: MRI faraday's cage

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Title: العنوان:

ATOMIC STRUCTURE

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1.6 Atomic Structure

All things are made of atoms, including the human body. Atoms are very small. Half a million lined up together are narrower than a human hair. Atoms are organized in molecules, which are two or more atoms arranged together. The most abundant atom in the body is hydrogen. This is most commonly found in molecules of water (where two hydrogen atoms are arranged with one oxygen atom, H_2O) and fat (where hydrogen atoms are arranged with carbon and oxygen atoms; the number of each depends on the type of fat).

The atom consists of a central nucleus and orbiting electrons. The nucleus is very small, one millionth of a billionth of the total volume of an atom, but it contains the entire atom's mass. This mass comes mainly from particles called nucleons, which are subdivided into protons and neutrons. Atoms are characterized in two ways. The atomic number is the sum of the protons in the nucleus. This number gives an atom its chemical identity. The mass number is the sum of the protons and neutrons in the nucleus. The number of neutrons and protons in a nucleus are usually balanced so that the mass number is an even number. In some atoms, however, there are slightly more or fewer neutrons than protons. Atoms of elements with the same number of protons but a different number of neutrons are called isotopes. Nuclei with an odd mass number (a different number of protons to neutrons) are important in MRI (see later).

Electrons are particles that spin around the nucleus. Traditionally this is thought of as being analogous to planets orbiting around the sun. In reality, electrons exist around the nucleus in a cloud; the outermost dimension of the cloud is the edge of the atom. The position of an electron in the cloud is not predictable as it depends on the energy of an individual electron at any moment in time (physicists call this Heisenberg's Uncertainty Principle). The number of electrons, however, is usually the same as the number of protons in the nucleus.

Protons have a positive electrical charge; neutrons have no net charge and electrons are negatively charged. So, atoms are electrically stable if the number of negatively charged electrons equals the number of positively charged protons. This balance is sometimes altered by applying external energy to knock out electrons from the atom. This causes a deficit in the number of electrons compared with protons and causes electrical instability. Atoms, in which this has occurred, are called ions.

1.7 Magnetization

The earth electrically charged and spinning ball is floating in space. Quite happily: nothing to worry about. From our physics lessons in school we may remember that a rotating electrical charge creates a magnetic field. And sure enough, the earth has a magnetic field, which we use to find our way from one place to another by means of a compass. The magnetic field strength of the earth is rather small: 30 T at the poles and 70T at the equator. (Tesla is unit for magnetic fields). In short we can establish that the earth is a giant spinning bar magnet, with a north and a south pole (Figure 9).

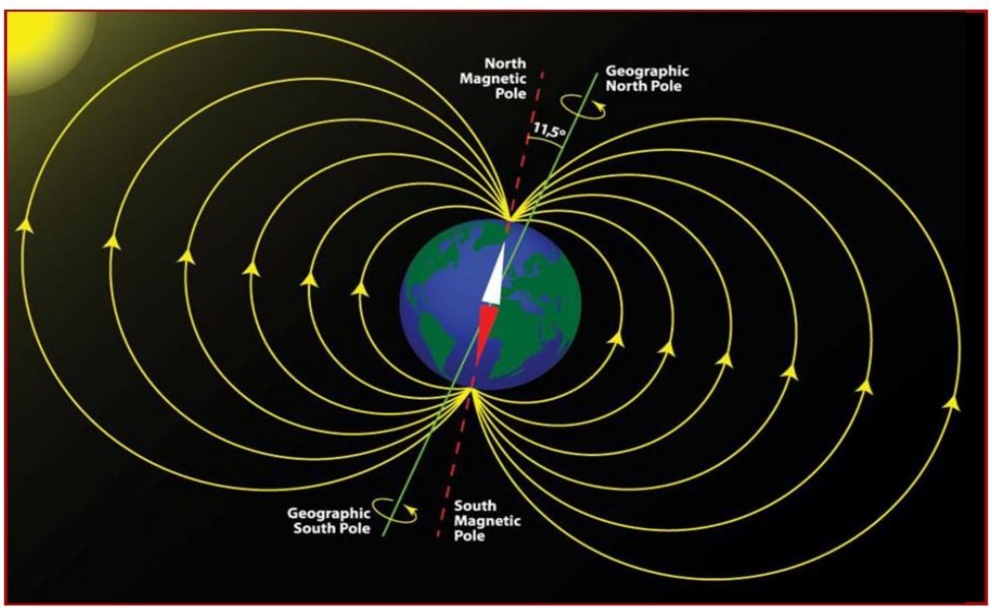

Figure 9: north and a south pole of earth

We have many in common with earth. If we take a bit from our body and we would put it under an electron microscope we can see things that look rather familiar. We see tiny little balls, which rotate around their own axes and also have an electrical charge and they have particles floating around it. This balls which we see are atoms. And atoms have everything to do with MRI, because we use them to generate our MR image. Another thing we have in common with earth is water. Our body consists of 80% water.

The moving charges give rise to magnetic fields. Consequently, we might expect that something that is charged and spinning would also possess a magnetic field. This is indeed true.

Well known in the chemistry that there are many different elements, to be precise there are 110 elements. Human body consists mainly of water "about 80% water". Water consists of one oxygen and two hydrogen atoms. Consider the simplest nucleus, which is hydrogen atom (the first element in the periodic table) has a nucleus contain one proton, and one electron orbital. This proton is electrically positive charged and it rotates around (spin) its axis.

Also, the hydrogen proton behaves as if it were a tiny bar magnet with a north and a south pole (Figure 10).

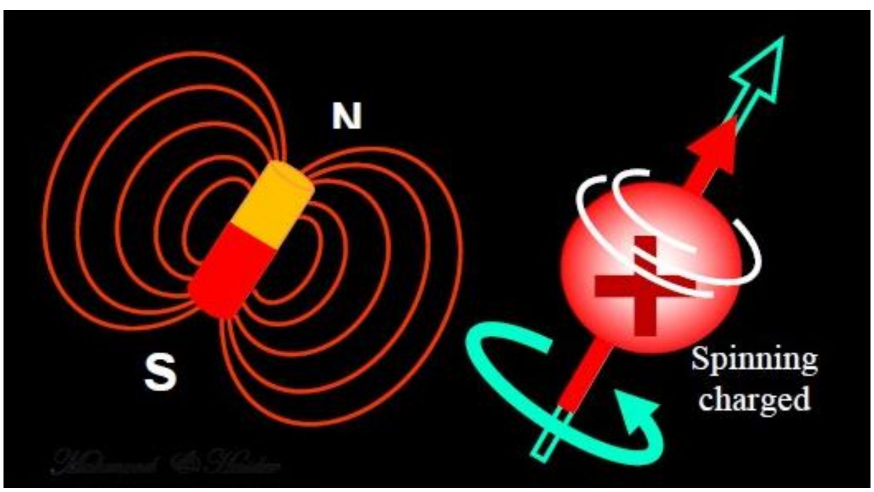

Figure 10: hydrogen proton. The positively charged hydrogen proton (+) spins about its axis and acts like a tiny magnet. *N* = north, *S* = south.

Hydrogen protons in the body thus act like many tiny magnets. The nucleus is said to be a **magnetic dipole**, and the name for its magnetism is **magnetic moment**. It is essential that there be a source of protons (protons in the nuclei of hydrogen atoms, which are associated with fat molecules and water) in order to form the MR signal.

We conclude from the above that there are two reasons for taking hydrogen as a source to form the MR signal or MR imaging source.

- First off all we have a lot of them in our body. Actually, it's the most abundant element we have.
- Secondly, in quantum physics there is a thing called "Gyro Magnetic Ratio". It is beyond the scope of this book what it represents; suffice to know that this ratio is different for each proton. It just so happens, that this gyro magnetic ratio for Hydrogen is the largest; 42.57 MHz/Tesla.

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Title: **العنوان:**

MAGNETIC MOMENTS

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1.8 Magnetic Moments

In most materials, such as soft tissue, these little magnetic moments are all oriented randomly (see figure 11). That is, if one nucleus has its spin and therefore its magnetic moment pointed up, there will be another nearby nucleus with its spin pointed down. Other magnetic moments will be oriented in various directions. This random orientation causes all the spins and magnetic moments to cancel, so that the **net magnetization** is zero. Net magnetization is symbolized by **M**.

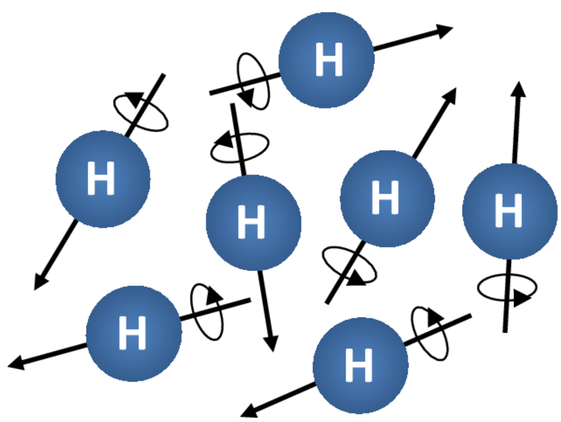


Figure 11: the magnetic moments of protons in all directions randomly**.**

If the patient however, is placed in a strong magnetic field, the magnetic moments will align themselves much as a compass needle aligns itself with the earth's magnetic field. Although all the magnetic moments are illustrated as being aligned in the same direction as the external magnetic field, in fact nearly as many align against the field as with it. It is a result of quantum mechanics that the moments must align either with the field or against it. A small excess of moments aligned with the field gives the patient a net magnetization, **M** as shown in figure 12.

The atoms in any material are in constant thermal motion, and thus the nuclei are being continually banged out of alignment. At any particular time however, slightly

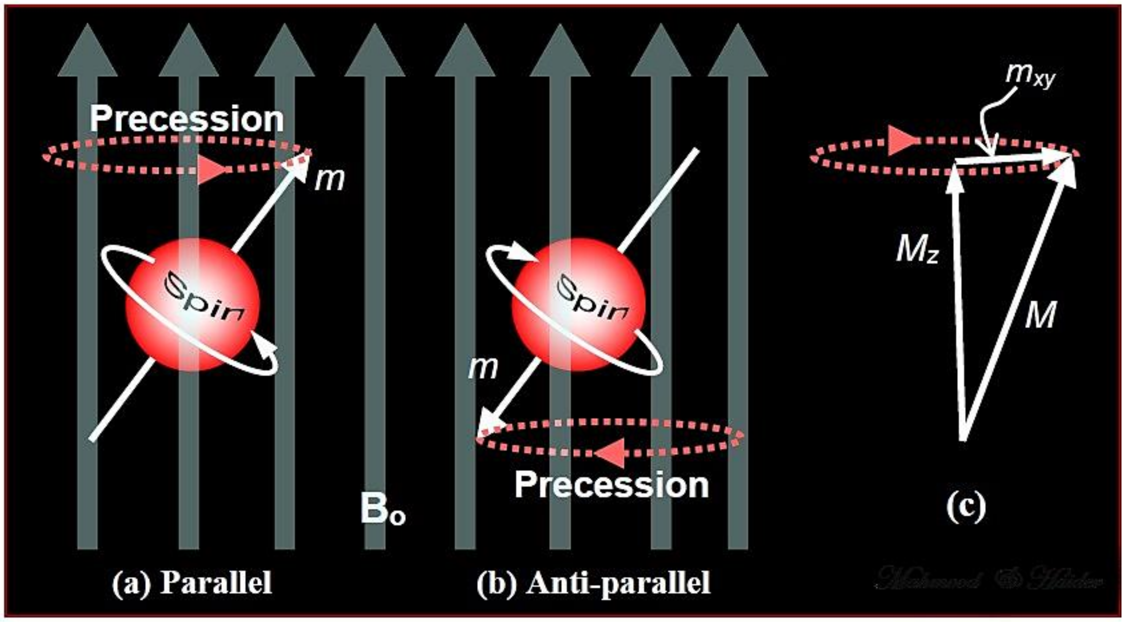


Figure 12: the hydrogen protons parallel or anti-parallel

more of the nuclei will align with the field than against it, creating net magnetization of the patient. The patient becomes a magnet.

For whom really wants to know, Hydrogen is not the only element we can use for MRI. In fact, any element, which has an odd number of particles in the nucleus, can be used. Some elements, which can be used, are: The protons in a molecule bunch of hydrogen a lot of tiny bar magnets spinning their own axes. As well know two north poles and two south poles of two magnets repel each other, while two poles of opposite sign attract each other. In human body these tiny bar magnets are ordered in such a way that the magnetic forces equalize. Human bodies are, magnetically speaking, in balance.

As we saw at the beginning of this chapter in the paragraph about the hardware, the magnets used in MR imaging can be in different field strengths. For example, the magnetic field strength of 1 Tesla magnet is ± 20000 times stronger than the Earth's gravitational field! This shows that we are working with the equipment to be potentially dangerous.

If the person placed in the MRI scanner some interesting things happen to the hydrogen protons:

1. They align with the magnetic field. This is done in two ways, **parallel** or **anti-parallel**.

2. They **process** or “**wobble**” around the direction of the external magnetic field (the z-axis) due to the magnetic momentum of the atom.

They process at a frequency called the Larmor frequency. Larmor frequency to its importance needs to be further explained. The Larmor frequency can be calculated from the following equation: $\omega_o = \gamma B_o$

Where ω_o = precessional or Larmor frequency (MHz)

γ = Gyro Magnetic Ratio (MHz/T)

B_o = Magnetic field strength (T)

Here we see the Gyro Magnetic Ratio and the Magnetic field strength, come together which the two discussed before. Highlights the importance of this equation is by the need to Larmor frequency to calculate the operating frequency in the magnetic resonance imaging system. For example, if the magnetic resonance imaging system 1.5 Tesla, then Larmor frequency or processional is: **42.57 × 1.5 = 63.855 MHz**

The precessional frequencies of 1.0T, 0.5T, 0.35T and 0.2T systems would work out to be 42.57 MHz, 21.285 MHz, 14.8995 MHz and 8.514 MHz respectively. When applied strong magnetic field of the scanner on protons, it could align with the field in two ways: parallel and anti-parallel.

Can also be called the two cases are low-energy state (parallel) and high energy state (anti-parallel). Distributions of protons for both states are not the same. To approximate the mind image can compare the protons just like a lot of people are lazy. They prefer to be in a low energy state. Protons aligned parallel which are low-energy State more than anti-parallel protons to the direction of the applied magnetic field which are a high-energy state (Figure 13). However, the difference between the two states is not large.

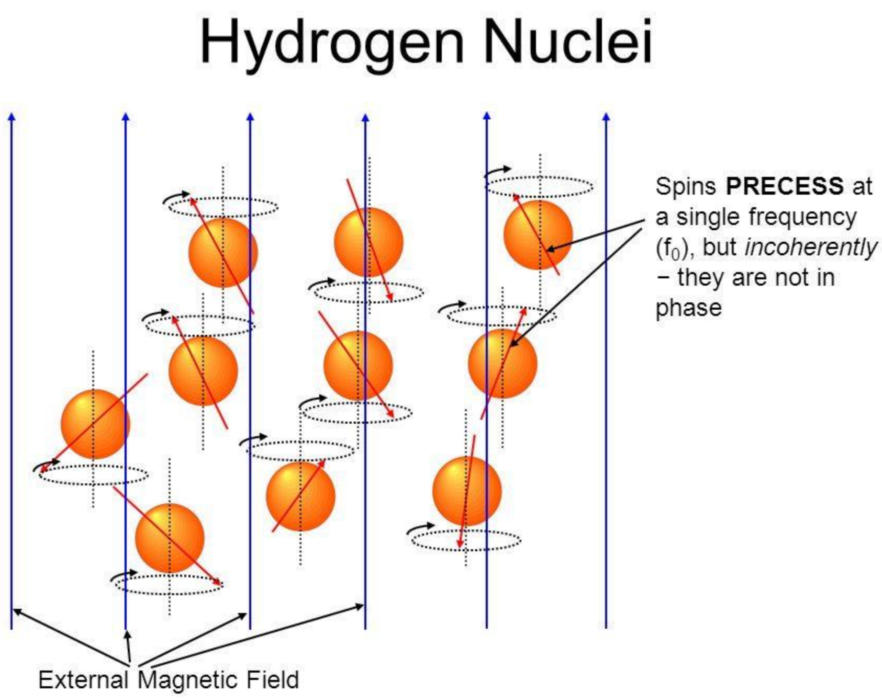


Figure 13: Hydrogen atoms and magnetic field

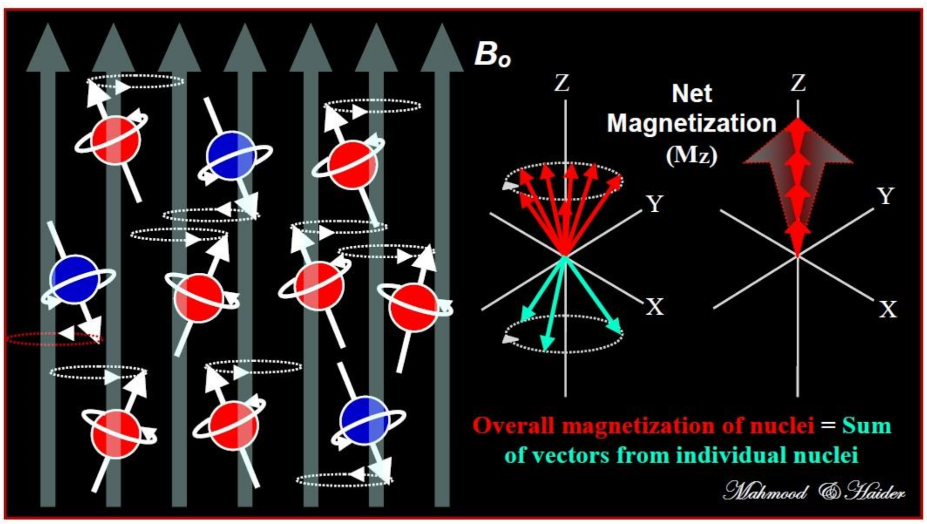
For example, the excess number of protons that aligned parallel or low energy state within a field 0.5T is only 3 per million (3 ppm = 3 parts per million), in a 1.0T system there are 6 per million in the system There 1.5T 9 per million. So, the excess number of protons is proportional with Bo. This is also the reason for the 1.5T systems make images better than systems with lower field strength.

The excess number of protons within a field 1.5T is only 9 per million which is don't seem very many, but in real life it adds up to quite a number. For example, if we calculated how many excess protons there are in a single voxel (volume element) at 1.5T.

- Assume a voxel is 2 x 2 x 5 mm = 0.02 ml
- Avogadro's Number says that there are 6.02 x 1023 molecules per mole.
- 1 mole of water weighs 18 grams (O16 + 2H1), has 2 moles of Hydrogen and fills 18 ml, so…………
- 1 voxel of water has 2 x 6.02 x 10^{23} x 0.02 / 18 = 1.338 x 10^{21} total protons
- The total number of excess protons = $(1.338\times10^{21}\times9)/2\times10^{6}$ $=6.02\times10^{15}$ or 6 million billion protons

Eventually we can see that there is a net magnetization (the sum of all tiny magnetic fields of each proton) pointing in the same direction as the system's magnetic field. Now, if like us, **net magnetization** using an easy by **vector** in order to see what is happening with them in MRI. A vector (the red arrow in the Figure 14) has a direction and a force. We imagine a frame of rotation, which is a set of axes called X, Y and Z. The Z-axis is always pointing in the direction of the main magnetic field, while X and Y are pointing at right angles from Z. Here we see the (red) net magnetization vector pointing in the same direction as the Z-axis. The net magnetization is now called M or longitudinal magnetization.

Figure 14: Direction and a force of net magnetization



To obtain an image from a patient it is not enough to put him into the magnet. We have to do a little bit more than that. What we also have to do is discussed in the following pages. The following steps can be divided into Excitation, Relaxation, Acquisition, Computing and Display. Before that, should understand the meaning of some of the important expressions: an **in-Phase** and **diphase**

الجامعة التقنية الوسطى
كلية التقنيات الصحية والطبية / بغداد
قسم : تقنيات الاشعة المادة: الفيزياء الشعاعية / الرنين المغناطيسي المرحلة: الثالثة

Title: **العنوان:**

RF PULSE

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قسم تقنيات الاشعة

Posttest: **الاختبار البعدي:**

References: **المصادر:**

Thayalan, K., and Ramamoorthy Ravichandran. *The physics of radiology and imaging*. JP Medical Ltd, 2014.

1.9 In-Phase and Diphase

To explain first what do we mean by **Phase**?

In Figure 15 we see two wheels with an arrow. (a) The wheels rotate in the same speed and in the same angle. The arrows will therefore point in the same direction at any time. Say the wheels rotate in the same phase (**in-Phase**). Another two wheels (b) with different angle therefore, we say out of phase (**de-phase**).

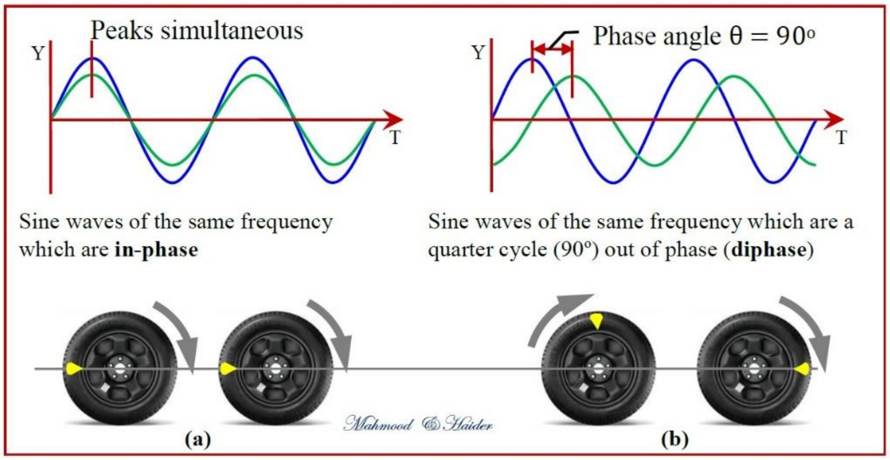

Figure 15: (a) two wheels are rotating **in-Phase** (b) two wheels are rotating **de-phase**

1.10 RF Pulse

As mentioned previously in section 1.8, if the person placed in the MRI scanner, the first thing which happens is the precession of spins around the direction of the external magnetic field (the z-axis). Now what happen, if another magnetic field is temporarily switched on in a different direction ***(let's say in the direction of the x- or y-axis)?***

Precession will occur around the direction of that magnetic field also.

If the second applied magnetic field is static, the resultant movement of the net magnetization of a spin isochromatic will be a complicated motion due to precession from the **two static fields**. However, if the second magnetic field which is temporarily applied is oscillating with the frequency of precession of the precessing spins a simple rotation of the net magnetization vector results. (Rotation of magnetization into the x-y-plane is a "90° pulse". The dephasing of the components of magnetization in the x-y-plane starts to occur straight away, as does the re-growth of magnetization in the z-direction as shown in next section).

1.11 Excitation

Before the system starts to be acquire the data must be perform a quick measurement (also, called pre-scan) to determine the frequency of protons which are spinning (Larmor frequency). Selecting this frequency is important because it uses the system for the next step.

Once the Larmor frequency is determined the system will start the acquisition. For now, we only send a radio frequency (RF) pulse (An RF pulse is a magnetic field, the direction of which is oscillating at the Larmor frequency) into the patient and we look at what happens.

The oscillating magnetic field at the Larmor frequency is switched on for a very small amount of time (a few milliseconds) to achieve such a rotation. This magnetic field is called an RF pulse; it is short (a burst or pulse) and the Larmor frequency for MRI is in the radio frequency range (tens of MHz). This process is sometimes called RF excitation of the spin system. Different amounts of rotation can be achieved by applying the oscillating magnetic field for different durations.

To understand it more deeply can through the following example:

Let us assume we work with a 1.5 Tesla system. The center or operating frequency of the system is 63.855 MHz (Calculated using the Larmor equation: $\omega_o=\gamma B_o$). In order to manipulate the net magnetization, we will therefore have to send a Radio Frequency (RF) pulse with a frequency that matches the center frequency of the system: 63.855 MHz This is where the Resonance comes from in the name Magnetic Resonance Imaging. Resonance you know from the opera singer who sings a high note and the crystal glass shatters to pieces. MRI works with the same principle.

Only protons that spin with the same frequency as the RF pulse will respond to that RF pulse.

If we would send an RF pulse with a different frequency, let's say 59.347 MHz, nothing would happen. Therefore, by sending an RF pulse at the **Larmor Frequency**, with certain strength (amplitude) and for a certain period of time it is possible to rotate the **net magnetization** into a plane perpendicular to the Z-axis, in this case the X-Y plane (Figure 16).

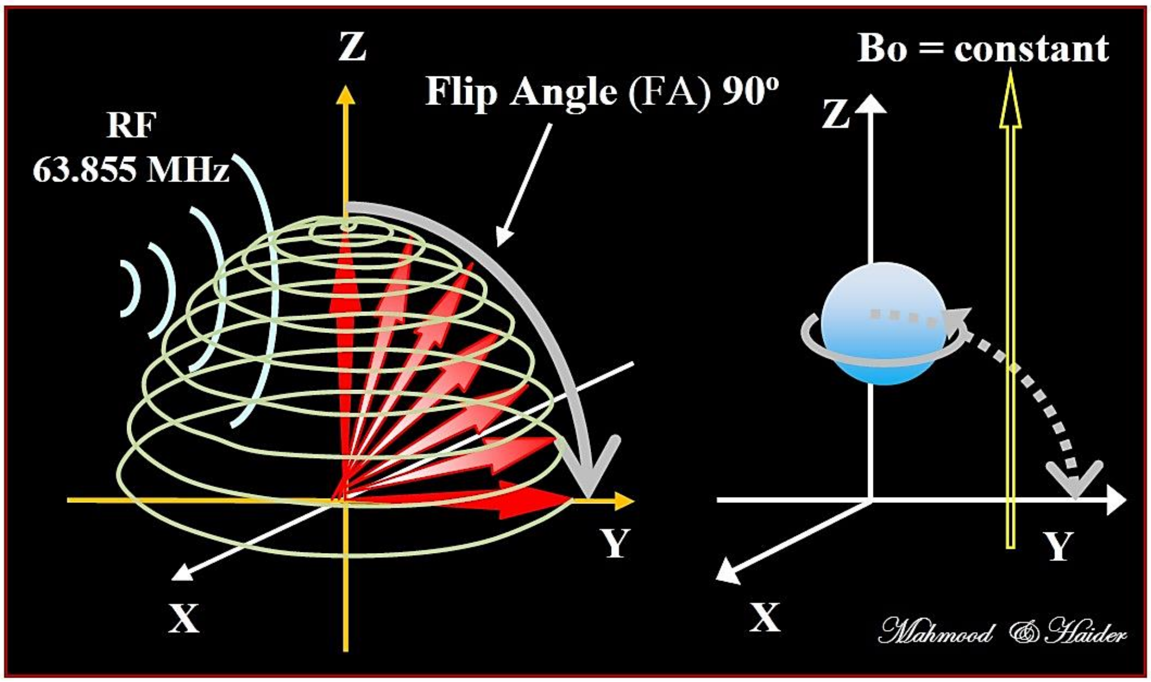

Figure 16: RF pulse at the Larmor Frequency, it is possible to rotate the net magnetization into a plane perpendicular to the Z-axis.

(**Note** how by use of the **vectors** facilitated imagine what is happening, without the use of vectors would be quite impossible that this event draws).

We just "flipped" the net magnetization 90º. Later we will see that there is a parameter in our pulse sequence, called the **Flip Angle** (FA), which indicates the number of degrees we rotate the net magnetization. It is possible to flip the net magnetization any degree in the range from 1^o to 180^o. For now, we only use an FA of 90^o. This process is called **excitation**.

1.12 Relaxation

Now it becomes interesting. If the net magnetization rotated 90 degrees in x-y plane and this means the same thing if we say that the protons raised to a **higher energy state**. This occurs because the protons absorbed energy from the **RF pulse**. This is called the perturbation that protons do not "like or want" continue in high energy situation "**excitation**" they tend to return to the normal or low energy situation "**equilibrium**". This can be compared with the abnormal situation in the case of walking on your hands, this is possible, but you do not want to continue this case for a long time and you inevitably you prefer the natural state is walking on your feet. A general principle of thermodynamics is that every system seeks its lowest energy level. The same thing for the protons, and they prefer the lineup with the main magnetic field or, in other words, they would be in a low power state.

The **relaxation** means the return of a perturbed system into the original situation "equilibrium" and each relaxation process can be characterization by a relaxation time. The relaxation process can be divided into two parts: **T1** and **T2 relaxation**.

Title: **العنوان:**

EXCITATION AND RELAXATION

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References: **المصادر:**

Thayalan, K., and Ramamoorthy Ravichandran. *The physics of radiology and imaging*. JP Medical Ltd, 2014.

1.12.1 T_1 Relaxations

T1 is spin-lattice relaxation time which relates to the recovery of the magnetization along z direction **after RF pulse**. We can say that this as the time it takes tissue to recover from an RF pulse so you can give another pulse and still get signal.

T_1 is called the **spin-lattice relaxation time** because it refers to the time it takes for the **spins** to give the energy, they obtained from the RF pulse back to the surrounding tissue (**lattice**) in order to go back to their equilibrium state. T_1 relaxation describes what happens in the Z direction. So, after a little while, the situation is exactly as before we sent an RF pulse into the patient. In other words,

Immediately after the 90° pulse, the magnetization $\mathbf{M}_{xy}$ precesses within the x–y plane, oscillating around the z-axis with all protons rotating **in-phase**. After the magnetization has been flipped 90° into the x–y plane, the RF pulse is **turned off**. Therefore, after the RF pulse is turned off, two things will occur:

1. The spins will go back to the lowest energy state.
2. The spins will get **out of phase** with each other.

The Protons are return to its original situation "equilibrium" by the releasing the absorbed energy in the form of "very little" warmth and RF waves. That means, in principle the net magnetization rotates back to align itself with the Z-axis. After the stops of the RF excitation pulse, the net magnetization will re-grow along the Z-axis, while emitting radio-frequency waves (Figure 18).

T_1 relaxation describes what happens in the Z direction. So, after a little while, the situation is exactly as before we sent an RF pulse into the patient. T_1 relaxation is also known as Spin-Lattice relaxation, because the energy is released to the surrounding tissue (lattice). So far, so good! This process is relatively easy to understand because one can, somehow, picture this in one's mind.

T_1 Relaxation Curves

T_1 relaxation happens to the protons in the volume that experienced the 90o-excitation pulse. However, not all the protons are bound in their molecules in the same way. This is different for each tissue. One 1H atom may be bound very tight, such as in fat tissue, while the other has a much looser bond, such as in water. Tightly bound protons will release their energy much quicker to their surroundings than protons, which are bound loosely. The rate at which they release their energy is therefore different. The rate of T_1 relaxation can be depicted as shown in Figure 19.

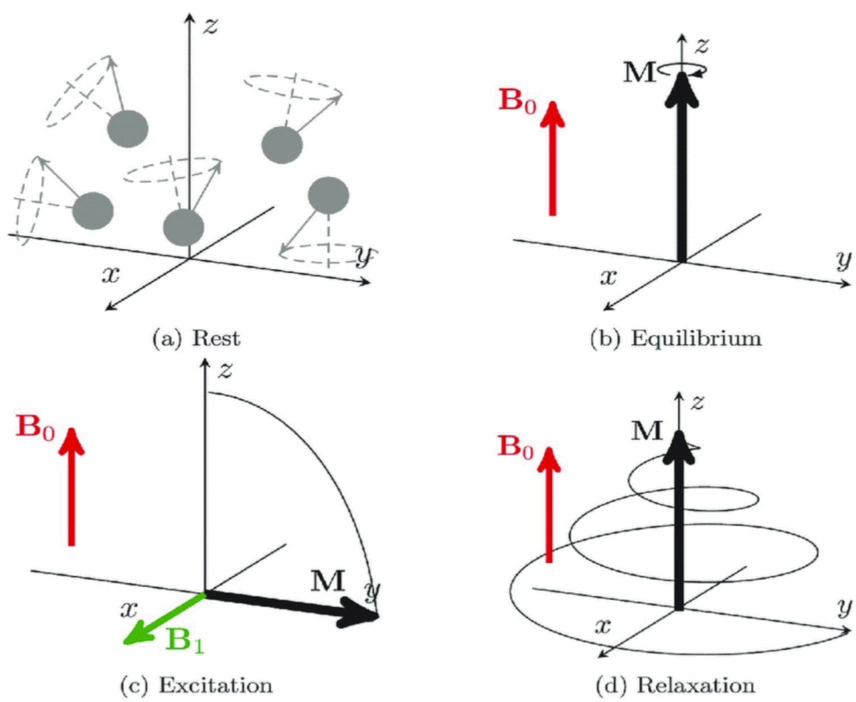


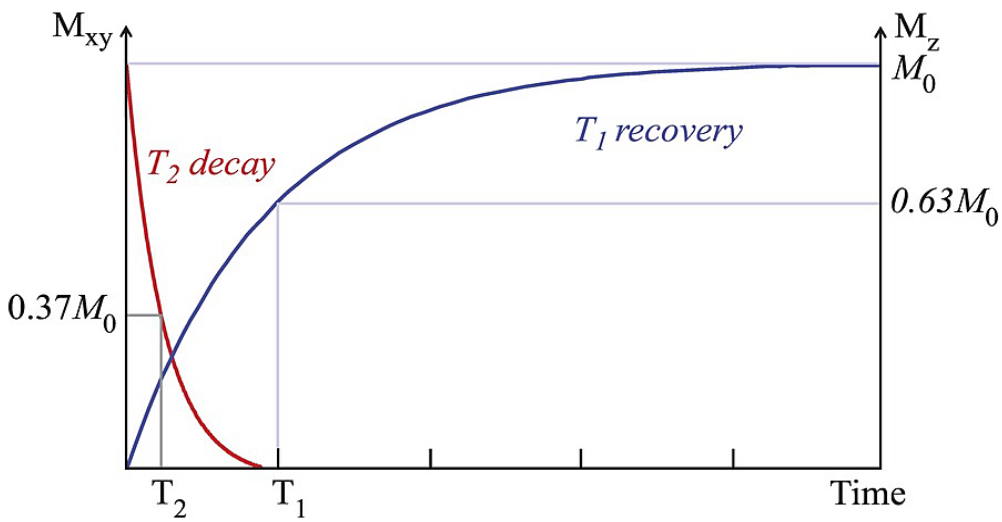
Figure 18: The net magnetization will re-grow along the Z-axis after the stops of the RF excitation pulse

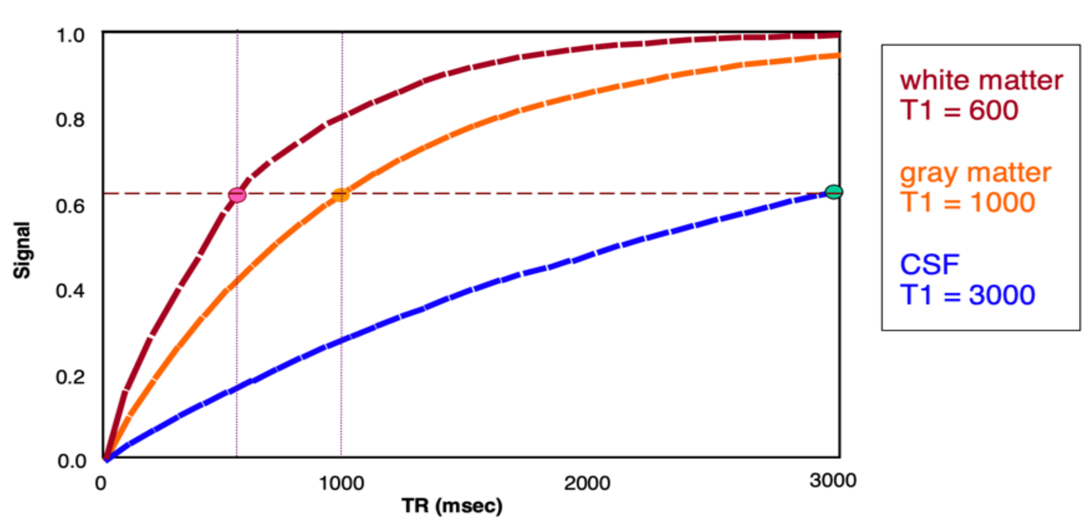
The curve shows at time = 0 that there is no magnetization in the Z-direction right after the F-pulse. But immediately the MZ starts to recover along the Z-axis. T_1 relaxation is a time constant. [T_1 is defined as the time it takes for the longitudinal magnetization (MZ) to reach 63 % of the original magnetization]. In other words, magnetization (MZ) be at the beginning and before sending a pulse on the Z-axis with the maximum value (100%), after sending a pulse, MZ go down to zero on the Z-axis and a full appear in the xy plane, (i.e., 90 degrees turn spiral path as shown in figure 18). This means that the protons are in an excited state. Immediately after cutting-off the pulse, begin declining in the xy plane and at the same time grow on the Z-axis to reach 63% of its original value with a time called T_1, (i.e., that the protons begin to return to equilibrium or primary status) according to the following equation:

$$M_z = M_o\,(1- e^{-t/T1})$$

For example, $t = T_1$ and $t = 5T_1$, answers will be ~ 63%, ~ 99% respectively.

A similar curve can be drawn for each tissue as shown in figure 20 which illustrates four tissues found in the head. Each tissue will release energy (relax) at a different rate and that's why MRI has such good contrast resolution.





1.12.2 T_2

Relaxations

As I mentioned before, the relaxation process is divided into two parts. The second part, relax T_2, is a bit more complicated. We have found that students in general and even some radiologists have difficulties in understanding T_2 in addition to understanding the relationship between T_1 and T_2.

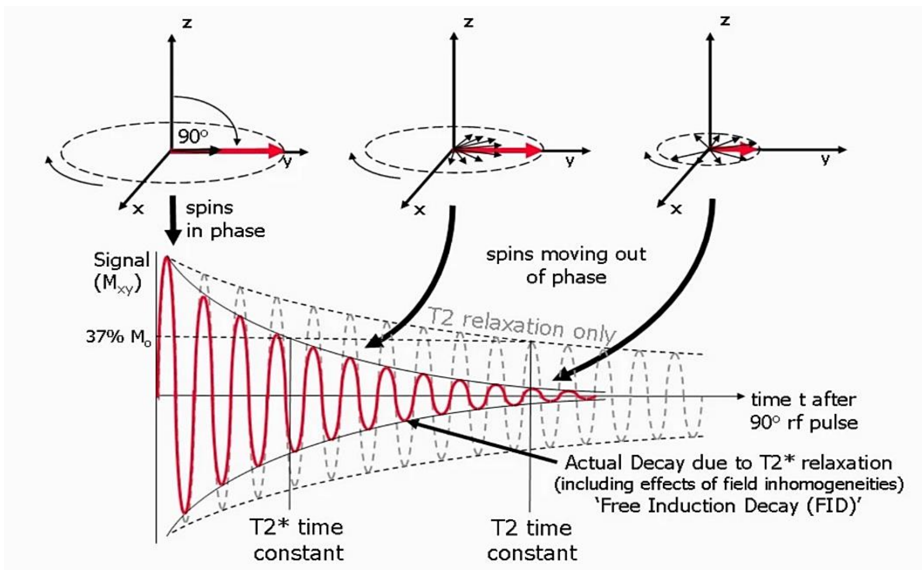
First of all, it is very important to realize that T_1 and T_2 relaxation are **two independent processes**. **The one has nothing to do with the other**. The only thing they have in common is that both processes happen simultaneously. T_1 relaxation describes what happens in the Z direction, while T_2 relaxation describes what happens in the X-Y plane. That's why they have nothing to do with one another.

Let's go back one step and have a look at the **net magnetization vector** before we apply the 90° RF pulse. The **net magnetization vector** is the sum of all the small magnetic fields of the protons, which are aligned along the Z-axis.

Each individual proton is spinning around its own axis. Although they may be rotating with the same speed, they are not spinning **in-phase** or, in other words, there is no **phase coherence**. The arrows of the two wheels from the previous example would point in different directions. When we apply the 90° RF pulse something interesting happens. Apart from flipping the magnetization into the X-Y plane, the protons will also start spinning **in-phase**!

So, right after the 90o RF pulse the net magnetization vector (now called **transverse magnetization**) is rotating in the X-Y plane around the Z-axis at the Larmor frequency (Figure 21).

That is transverse magnetization formed by tilting the longitudinal magnetization into the transverse plane by using a radiofrequency pulse. The transverse magnetization induces an MR signal in the radiofrequency coil immediately after



its formation, it has a maximum magnitude, and all of the protons are in phase. Therefore, the vectors all point in the same direction because they are **in-Phase**. However, they don't stay like this. The transverse magnetization starts decreasing in magnitude immediately as protons start going out of phase. This process of **de-phasing** and reduction in the amount of transverse magnetization is called transverse relaxation.

This is similar to a group of soldiers walking one behind the other in a similar pattern (**in- phase**). If someone stumbled resulting in a state of mini chaos with other soldiers who are walking and then walk change in different directions: this soldier got out-of-Phase or he were **(de-phasing)**.

A similar situation happens with the vectors in MRI. Remember that each proton can be thought of as a tiny bar magnet with a north and a south pole. And two poles of the same sign repel each other. Because the magnetic fields of each vector are influenced by one another the situation will occur that one vector is slowed down while the other vector might speed up. The vectors will rotate at different speeds and therefore they are not able to point into the same direction anymore: they will start to de-phase. At first the amount of de-phasing will be small (Figure 21), but quickly that will increase until there is no more phase coherence left: there is not one vector pointing in the same direction anymore.

In the meanwhile, the whole lot is still rotating around the Z-axis in the X-Y plane (Figure 21). A characteristic time representing the decay of the signal by 1/e, or 37%, is called the T_2 relaxation time. $1/T_2$ is referred to as the transverse relaxation rate. This process of getting from a total **in-phase** situation to a total **out-of-phase** situation is called **T_2 relaxation**.

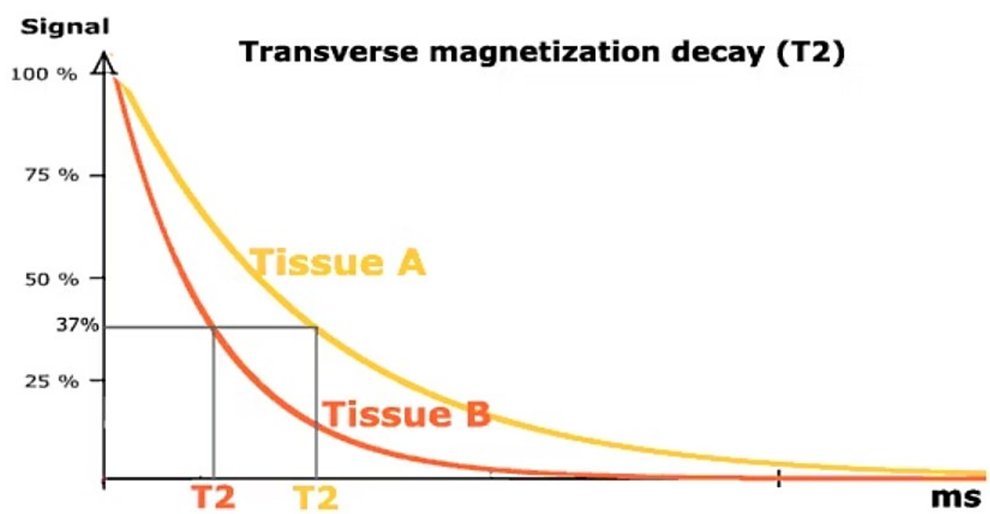
T_2 Relaxation Curves

Just like T_1 relaxation, T_2 relaxation does not happen at once. Again, it depends on how the Hydrogen proton is bound in its molecule and that again is different for each tissue.

Right after the 90^o RF-pulse all the magnetization is "flipped" into the XY-plane. The net magnetization changes name and is now called MXY. At time = 0 all spins are in-phase, but immediately start to de-phase. T_2 relaxation is also a time constant. T_2 is defined as the time it takes for the spins to de-phase to 37% of the original value. Immediately after cutting-off the pulse, begin declining in the xy plane, according to the following equation:

$$M_{xy} = M_o e^{-t/T2}$$

The rate of **de-phasing** is different for each tissue. Fat tissue will de-phase quickly, while water will de-phase much slower, which can be shown in figure 22. One more remark about T_2: it happens much faster than T_1 relaxation. T_2 relaxation happens in tens of milliseconds, while T_1 can take up to seconds. T_2 relaxation is also called spin–spin relaxation because it describes interactions between protons in their immediate surroundings (molecules).



T2* Relaxation

All relaxation mechanisms mentioned so far are heavily influenced by temperature and molecular environment. Transverse relaxation is the result of random interactions at the atomic and molecular levels. Transverse relaxation is primarily related to the intrinsic field caused by adjacent protons (spins) and hence is called spin-spin relaxation. Transverse relaxation causes irreversible de-phasing of the transverse magnetization.

By contrast, the so-called T_2* relaxation (as a variant of T_2) is a result of dephasing processes due to an inhomogeneous magnet field which can be minimized by manual justification ("shimming"). Since T_2* is usually much smaller than T_2, the signal decay of an FID is almost completely caused by T_2* effects. In general, $T_1>T_2>T_2^*$.

Remember this:

- T_1 and T_2 relaxation are two independent processes, which happen simultaneously.
- T_1 happens along the Z-axis; T_2 happens in the X-Y plane.
- T_2 is much **quicker** than T_1

When both relaxation processes are finished the net magnetization vector is aligned with the main magnetic field (B_0) again and the protons are spinning Out-Of-Phase; the situation before we transmitted the 90^o RF-pulse.

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ACQUISITION

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References: المصادر:

Thayalan, K., and Ramamoorthy Ravichandran. *The physics of radiology and imaging*. JP Medical Ltd, 2014.

1.13 Acquisition

During the relaxation processes the spins shed their excess energy, which they acquired from the 90o RF pulse, in the shape of radio frequency waves. In order to produce an image, we need to pick up these waves before they disappear into space. This can be done with a **Receive coil**. The receive coil can be the same as the **Transmit coil** or a different one. An interesting, but ever so important, fact is the position of the receive coil.

The receive coil must be positioned at right angles to the main magnetic field (B_0). Failing to do so will result in an image without signal. This is why: if we open up a coil, we see it is basically nothing but a loop of copper wire. When a magnetic field goes through the loop, a current is induced (Figure 23).

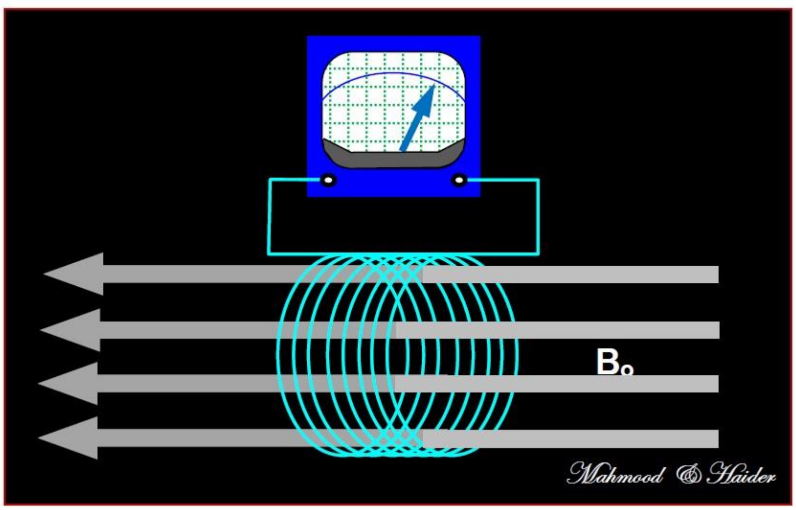
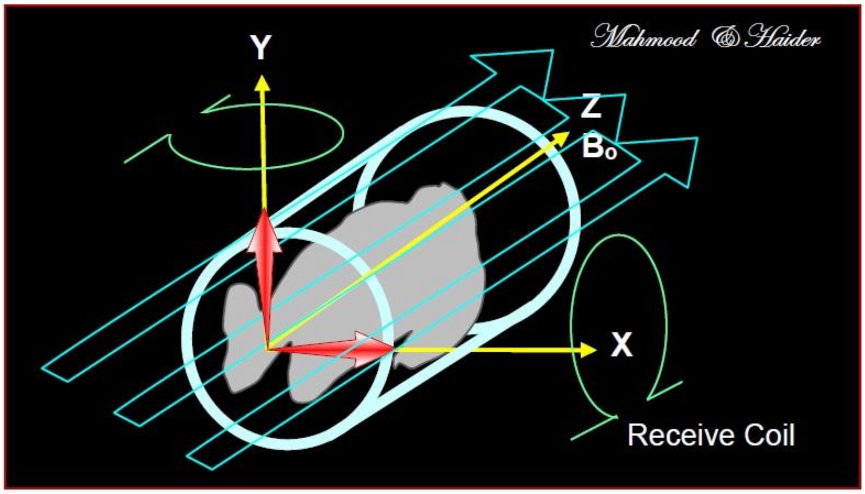


Figure 23: A magnetic field goes through the loop, a current is induced

Bo is a very strong magnetic field; much stronger than the RF signal we are about to receive. That means if we position the coil such that B_0 goes through the coil an enormous current is induced, and the tiny current induced by the RF wave is overwhelmed. We will only see a lot of speckles (called: noise) in our image. Therefore, we have to make sure that the receive coil is positioned in such a way that

B0 can't go through the coil. The only way to achieve this is to position the receive coil at right angles to B_0 as shown in Figure 9.24.

It is quite interesting to try this for yourself with your scanner. Just make a series of scans where you position the receive coil at different angles. Start with the coil at a right angle with B0, and then turn it a bit such that B0 is allowed to run through the coil. Next turn it a bit further until B_0 runs entirely through the coil. You will see your image degrade very quickly. At some stage the system is probably not able to "tune" the coil anymore and won't be able to make a scan.



Remember this:

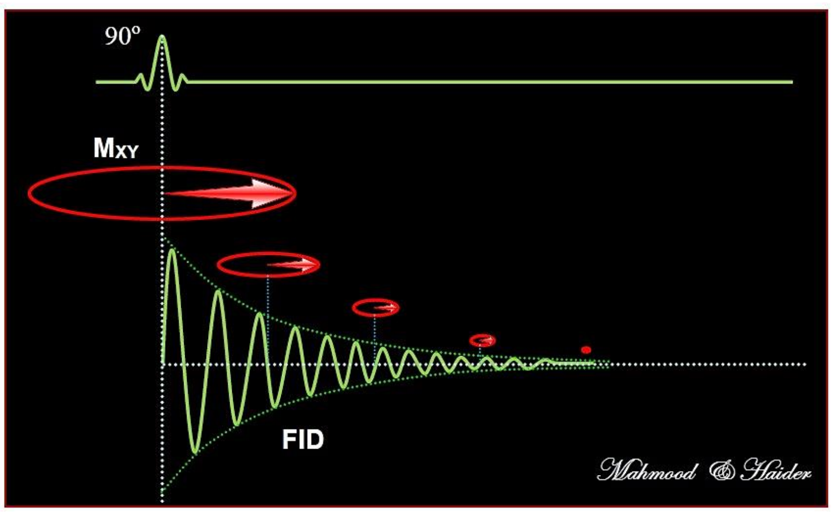
- The only proper way to position the receive coil is at right angles to B_o.

Note: Many coils are specifically designed for a certain body part. Take for instance the Head coil; if you fix the coil on the scanner table it seems that B_0 runs through the coil. This is only 'optical Illusion'. The coil is designed such that the loops of copper wire, which make up the coil, are at right angles to B_0. Designing a coil for a bore type magnet where B_0 runs through the length of the body is exceptionally difficult. If you open up a Head coil you'll see probably two copper wires, which are saddle shaped and positioned at right angles to one another. In order to receive enough signal, there are two coils, because saddle shaped coils are relatively inefficient.

According to **Mr. Faraday** a Radio Frequency wave has an electric and a magnetic component, which are at right angles from one another, have a 90^o phase difference and both move in the same direction with the speed of light.

It is the magnetic component in which we are interested because that induces the current in the receive coil.

Determine the positioning of the coils so that they form a right angle with B_0 means we can receive signals from processes only when the right angles be a between the coils and B_0 which happens to be T_2 relaxation. T_2 relaxation is a decaying process, which means phase coherence is strong in the beginning, but rapidly becomes less until there is no phase coherence left. Consequently, the signal that is received is strong in the beginning and quickly becomes weaker due to T_2 relaxation (Figure 25).

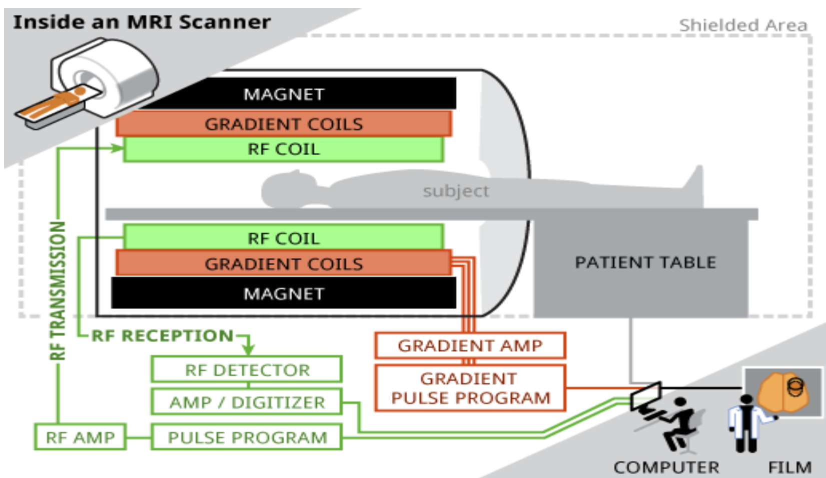


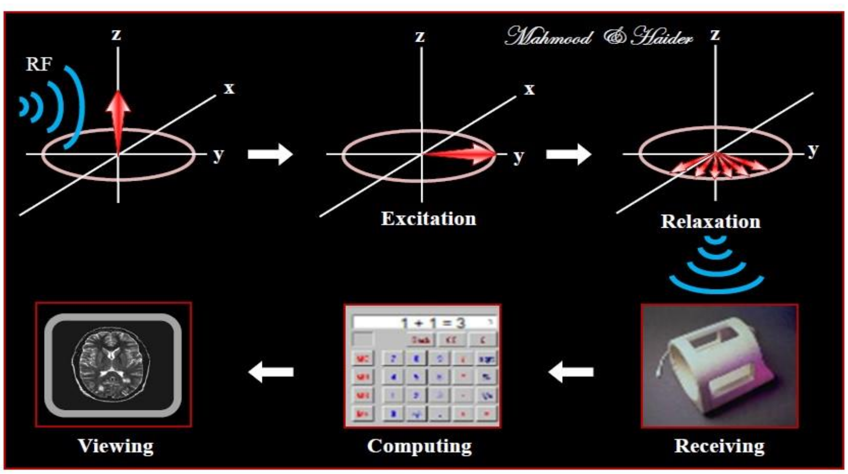
The signal is called: Free Induction Decay (FID). The FID is the signal we would receive in absence of any magnetic field. In the presence of a magnetic field T_2 decay goes much faster due to local (microscopic) magnetic field inhomogeneity and chemical shift, which are known as T_2* effects). The signal we receive is much shorter than T_2. The actual signal decays very rapidly; in ± 40 milliseconds it's reduced to practically zero. This poses a problem, as we will see later.

1.14 Computing and Display

In general, system of MRI consists of five major components: **magnet**, **gradient systems**, **RF coil system**, **receiver**, and **computer system**. Figure 26 shows the entire process graphically.

The received signal (Figure 27) is then fed into a computer; computer system is beyond the scope of your study.





So far, we learned how to work magnetic resonance imaging (MRI). Good that can be likened to that there is something happening and you at this time you see what is happening, but you did not participate. This is the time to get out and go a little deeper and participate what is happening.

1.15 Gradient Coils

As we explained previously to produce an image, you must stimulate the hydrogen nuclei in the body, and then determine the location of those nuclei within the body. These tasks are accomplished using the **gradient coil**.

To make it clear that more than during the following assumption: If we assume a completely homogeneous magnetic field (this ideal situation does not exist), then all the protons in the body will **spin** at the **Larmor frequency**. This also means that all protons when you return to equilibrium give the same signal. In this case we will not know whether the signal coming from the head or foot. So, you will not get a clear image.

The solution to our problem can be found in the characteristics of the RF-wave, which are:

Phase, Frequency and **Amplitude**

First, we will divide the body up to the volume elements, also known as: **voxels**.

- ✓ The protons within that voxel will emit RF wave with known **phase** and **frequency.**
- ✓ **Amplitude** of the signal depends on the number of protons in the voxel.

The answer to our problem is: **Gradient Coils**

The **gradient coils** are resistant type electromagnets, which enable us to create additional magnetic fields, which are, in a way, superimposed on the main magnetic field B_0. The gradient coils are used to spatially encode the positions of the MRI spins by varying the magnetic field linearly across the imaging volume such that the **Larmor frequency** varies as a function of position.

To achieve adequate image quality and frame rates, the gradient coils in the MRI imaging system must rapidly change the strong static magnetic field by approximately 5% in the area of interest. High-voltage (operating at a few kilovolts) and high-current (100s of amps) power electronics are required to drive these gradient coils.

To differentiate tissue types, the MRI systems analyze the magnitude of the received signals. Excited nuclei continue to emit a signal until the energy absorbed during the excitation phase has been released. The time constant of these exponentially decaying signals ranges from tens of milliseconds to over a second; the recovery time is a function of field strength and the type of tissue. It is the variations in this time constant that allow different tissue types to be identified.

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SIGNAL CODING

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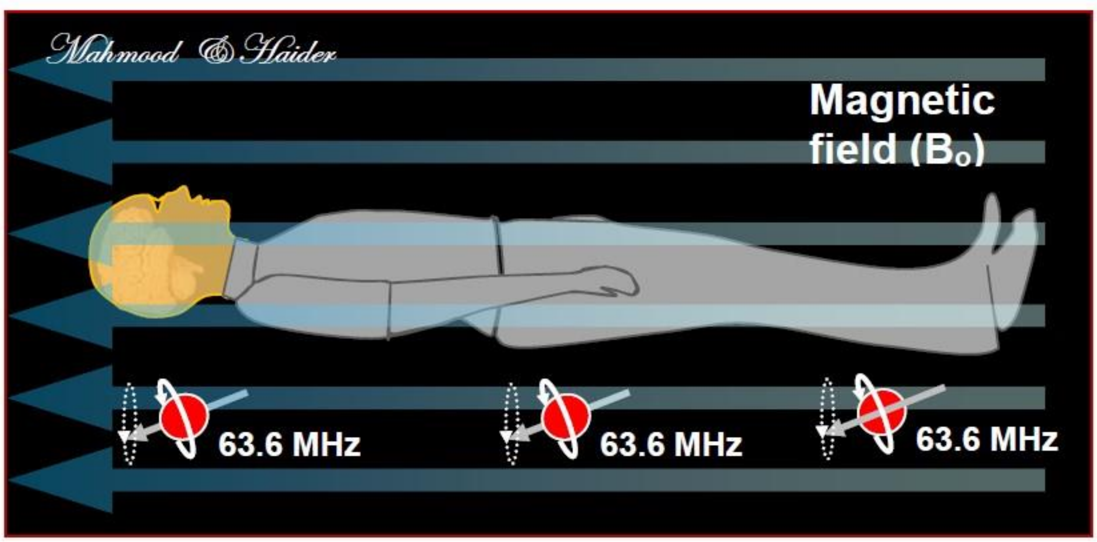
References: **المصادر:**

Thayalan, K., and Ramamoorthy Ravichandran. *The physics of radiology and imaging*. JP Medical Ltd, 2014.

1.16 Signal Coding

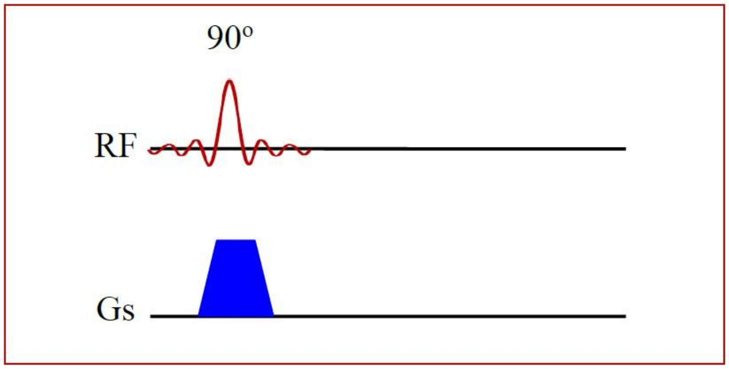
To explain this subject clearly and easy assimilation, suppose some of the assumptions: considering an axial image of the brain using a 1.5 Tesla magnet. Also, we work with a homogeneous magnetic field, which covers the whole body from head to toe. (This is quite different in reality, where there is only a homogenous sphere of 40 cm in diameter in the iso-centre (center of the MRI bore) of the magnet, but this assumption easy to the explanation and the idea explained). Just as important as the strength of the main magnet is its precision. The straightness of the magnetic lines within the center (or, as it is technically known, the iso-center) of the magnet needs to be near-perfect. This is known as homogeneity. Fluctuations (inhomogeneities in the field strength) within the scan region should be less than three parts per million (3 ppm). When we put a patient in the magnet, all the protons, from head to toe, align with B0. They spin at the Larmor frequency of 63.6 MHz (Figure 28).



If we use a 90° excitation RF-pulse to flip the magnetization into the x-y plane, then all the protons would react and return a signal. We would have no clue where the signal comes from: from head or toe.

1.16.1 Slice Encoding Gradient

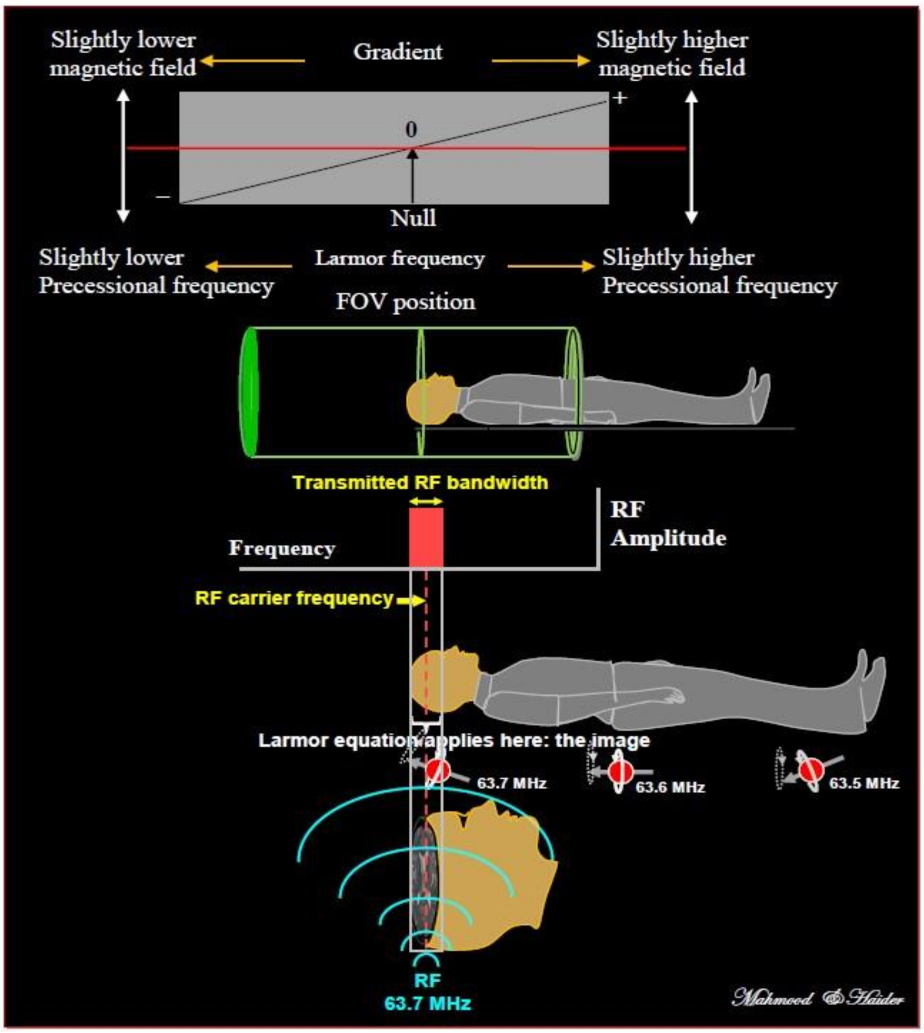
The magnetic field gradient (e.g., Z-gradient) is temporarily applied (Z-gradient is switched on) at the same time as the RF pulse (see figure 29).



This will generate an additional magnetic field in the Z-direction, which is superimposed on B_o. The indication $+G_z$ in Figure 29 means there is a slightly stronger Bo field in the head as there is in the iso-center of the magnet. A stronger B_o field means a higher Larmor frequency. Along the entire the slope of the gradient there is a different B_o field and consequently the protons spin at slightly different frequencies. Therefore, the protons in the head will spin slightly faster than the ones in the iso-center. The reverse goes for the protons in the feet. Figure 28 shows that the protons in the feet now spin at 63.5 MHz, the ones in the iso-center of the magnet still at 63.6 MHz and the ones in the head with 63.7 MHz

This means we can "pick out" the section which we want to excite by choosing the right frequency range of RF excitation pulse. The section which contains Larmor frequencies which match the frequencies of the oscillating magnetic field will respond. An MRI signal will be generated only from that section of the patient. This is called **Slice-Encoding** or **Slice-Selection**. Usually, the slice selection gradient is applied in the z-axis-the head-foot direction in the scanner. But because a gradient magnetic field may be applied in any orientation, slices may be acquired at literally any angle or orientation in the patient. This is one of strength MRI.

Now, if we apply an RF-pulse with a frequency of 63.7 MHz only the protons in a thin slice in the head will react because they are the only ones which spin with the same frequency. In this example G_z is the slice-encoding gradient. If we would stop here, this means that we receive the returned signal comes from the single slice in the head. That is, we have identified the **site** by using the **Z-gradient** (Gz). (Figure 30).



These frequencies are only used for this example; in reality the differences are much smaller.

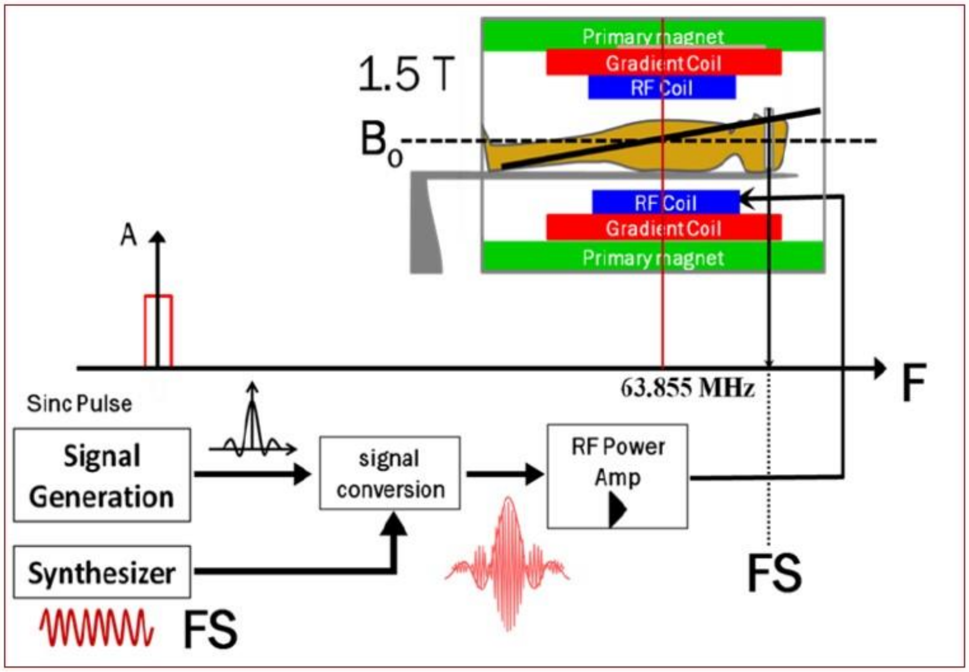
1.16.1.1 Slice Location

- The slice to be imaged is moved to the center of the scanner bore (the iso-center).

If there's time between imaging different slices, we can simply move the patient-table so that the section of interest within the patient is at the iso-center. This is preferable because placing the section of interest in the part of the main magnetic field which is most homogenous will give us better images.

The carrier frequency of the RF excitation pulse may be changed as shown in figure 31.

The carrier frequency of the transmitted RF pulse determines which spins along the patient will resonate (because they have a matching Larmor frequency). If multiple slices are to be acquired in quick sequence, the carrier frequency can be set to determine the location of the imaging slice in the patient.



1.16.1.2 Slice Thickness

Slice thickness helps get better resolution and finer detailed images. The slice thickness is governed by the following equation:

$$\text{Thickness} = BW_{trans} / \boldsymbol{\gamma_0} \boldsymbol{.}\ G_s$$

Where;

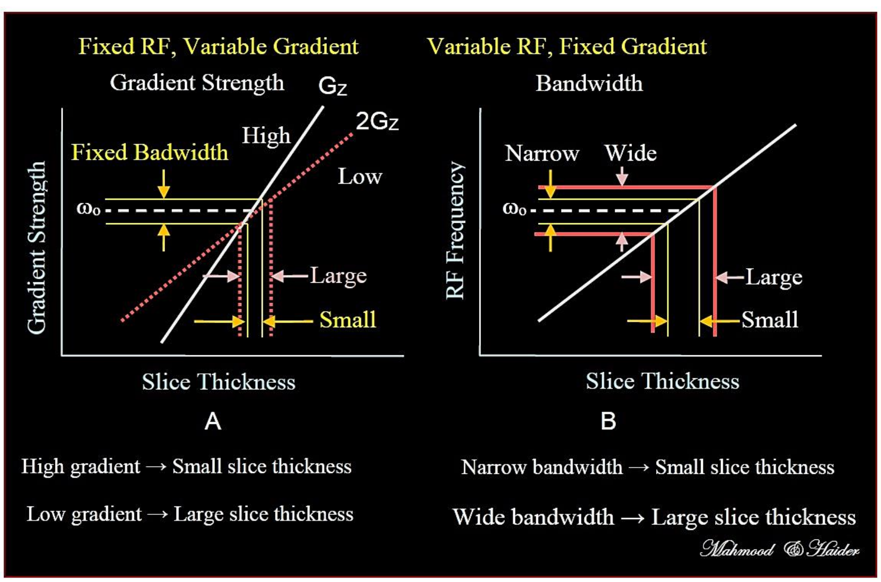
thk is the slice thickness,

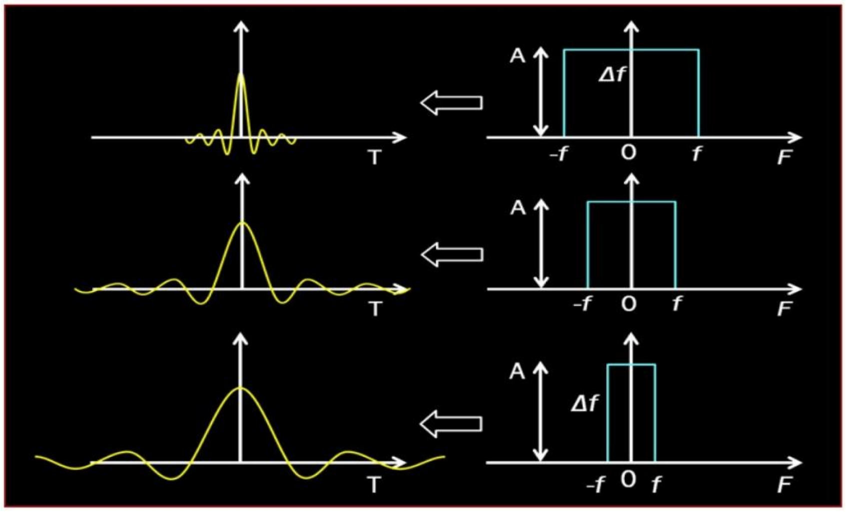
BWtrans is the transmitted RF bandwidth (the range of frequencies it covers),

$\boldsymbol{\gamma_0}$ is the gyromagnetic ratio and

Gs is the magnitude of the slice selection magnetic field gradient.

In Figure 32A show that varying the steepness of the gradient, while keeping the RF-pulse bandwidth the same. Alternatively, Figure 9.37B the steepness of the gradient is kept the same, while the bandwidth of the RF-pulse is varied. Can also change the slice thickness, the slice thickness may be reduced by either increasing the gradient of the magnetic field (dashed line in figure 32A) or by decreasing the RF pulse width, (or transmit bandwidth, figures 32B & 33). A thinner slice produces better anatomical detail, the partial volume effect being less, but it takes longer to excite.





In practice, the slice thickness is determined by a combination of both gradient steepness and RF-pulse bandwidth.

The total magnetic field at a position Z_{SS} (SS = slice selection) during application of G_Z is given by: $B_0+Z_{SS}\cdot G_Z$ and the spatially selective excitation energy or frequency, respectively, can be easily calculated using the Larmor equation.

A typical slice thickness is 2-10 mm. the RF pulse inevitably contains a certain amount of electromagnetic energy of frequencies slightly higher or lower than the intended bandwidth, thus mildly exciting tissues either side of the desired slice. To prevent this affecting the image slice, a gap (say 10% of the slice thickness) may be left between slices, although this is not necessary when the slices are interleaved.

- ❖ **For example,** for a 10 mm slice thickness using a gradient magnetic field strength of (10mmT/m), the transmitted RF pulse bandwidth would be about 4.3 kHz (using γ_0 = 42.58 MHz / T).

In order to get optimal image resolution, must be very thin slices with a high SNR. But whenever were thinner slices the noise was more, the SNR decreases and spatial resolution increase. Spatial Resolution is the ability to distinguish one structure from another. Conversely, increase of the slice thickness led to increase signal to noise ratio and reduces spatial resolution. Because the thicker slices result other problems such as an increase in partial volume effects.

Effects The poorer SNR of thin slices can be addressed for to some extent by increasing the number of acquisitions or by a longer TR. Yet this is accomplished only at the expense of the overall image **acquisition time** (the period of time required to collect the image data. This time does not include the time necessary to reconstruct the image. ADC - analog-to-digital converter) and reduces the cost efficiency of the MR imaging system. The slice thickness can be determined by: (1) the steepness of the slope of the gradient field (G_{ss}) and (2) the bandwidth of the 90^o RF-pulse.

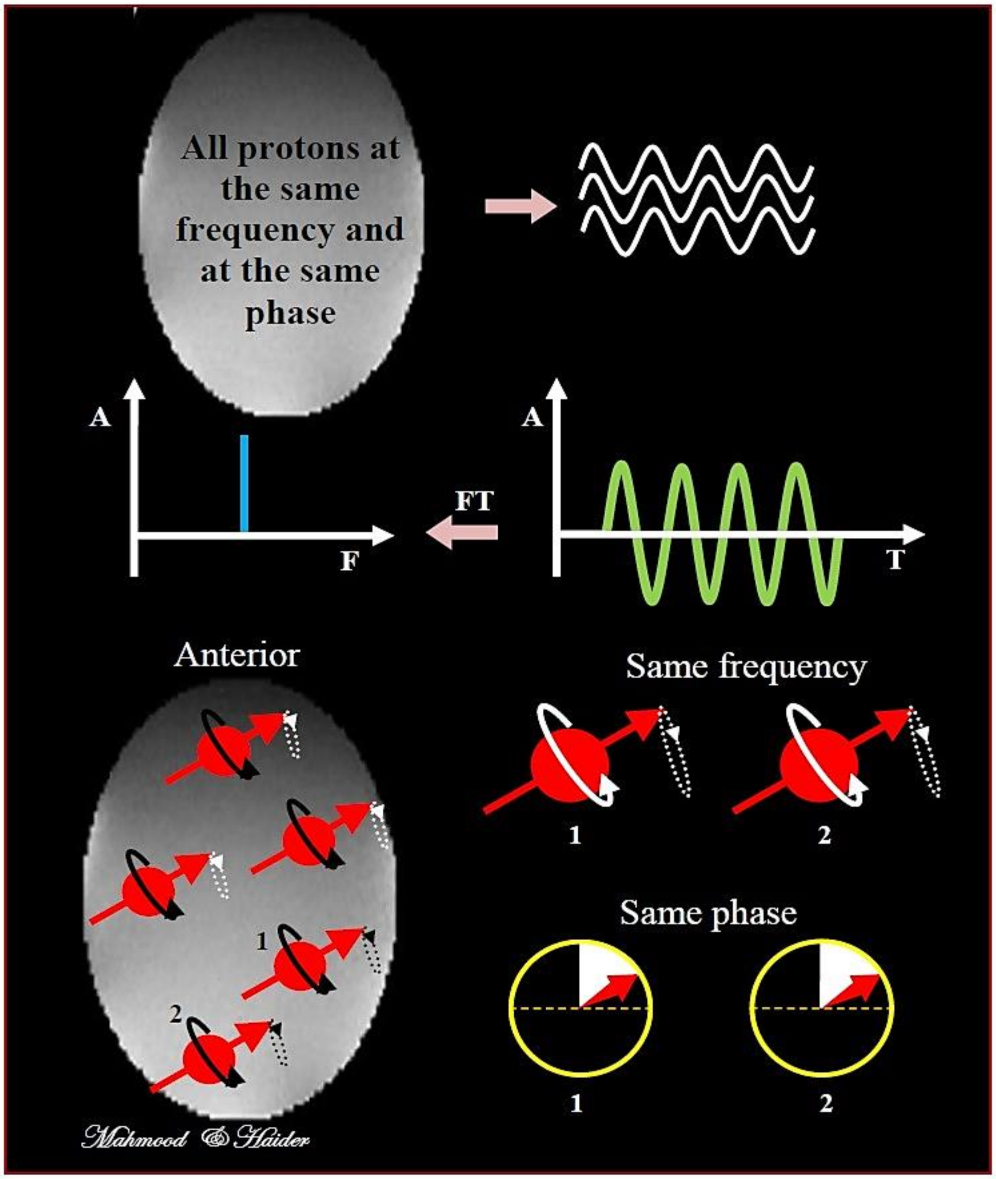
1.16.1.3 Receiver Bandwidth

The receiver bandwidth is the range of frequencies collected by an MR system during frequency encoding. The bandwidth is either set automatically or can be changed by the operator. A wide receiver bandwidth enables faster data acquisition and minimizes chemical shift artifacts but also reduces SNR as more noise is included. Halving the bandwidth improves SNR by about 30%. With a narrow bandwidth, on the other hand, there will be more chemical shift and motion artifacts and the number of slices that can be acquired for a given TR is limited.

Is this enough? Certainly not, and we will know immediately why.

Figure 34 shows the axial slice, which has just been created by the Gz gradient. If we take a closer look at proton 1 and 2 in this slice, we see that they both spin with the same frequency and have the same phase.

Within the slice there are still an awful lot of protons and we still don't know from where the signal is coming from within the slice. Whether it is comes from anterior, posterior, left or right. Further encoding is therefore required in order to allow us to pinpoint the exact origin of the signals.



Important Notes:

- The thickness of the slice can be changed by varying the steepness of the magnetic field gradient, or by changing the transmitted RF pulse bandwidth as will be discussed later in detail (section 9.8.2).
- The RF pulse and the magnetic field gradient have to apply together. This process may be depicted in a pulse sequence timing diagram.
- The shape of the RF excitation pulse **in time** is not a square (on/off) shape. This is because to excite a discrete range of frequencies (a slice) a **sine** shape pulse is used which can be seen by calculating sin(x)/x.

الجامعة التقنية الوسطى

كلية التقنيات الصحية والطبية / بغداد

قسم : تقنيات الاشعة **المادة: الفيزياء الشعاعية / الرنين المغناطيسي** **المرحلة: الثالثة**

Title: **العنوان:**

FREQUENCY ENCODING GRADIENT

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Target population: **الفئة المستهدفة:**

3rd year

طلبة المرحلة الثالثة

قسم تقنيات الاشعة

Posttest: **الاختبار البعدي:**

References: **المصادر:**

Thayalan, K., and Ramamoorthy Ravichandran. *The physics of radiology and imaging*. JP Medical Ltd, 2014.

1.16.2 Frequency Encoding Gradient

Figure 9.36 shows the axial slice, which has just been created by the Gz gradient. The protons in this slice have spin with the **same frequency** and have the **same phase**. Within the slice there are still an awful lot of protons and we still don't know from where the signal is coming from within the slice. Whether it is comes from anterior, posterior, left or right.

Further encoding is therefore required in order to allow us to pinpoint the exact origin of the signals. The frequency encoding gradient is a static gradient field, just like the slice selection magnetic field gradient. It does the same thing; it causes range of Larmor frequencies to exist in the direction in which it is applied (according to the Larmor equation).

To encode in the left – right direction the second, gradient (Gx) is switched on. This will create an additional gradient magnetic field in the left – right direction. we need now is to do one more encoding to determine whether the signal comes from the left, the center or the right side of the head. The protons on the left-hand side spin with a lower frequency than the ones on the right (Figure 35).

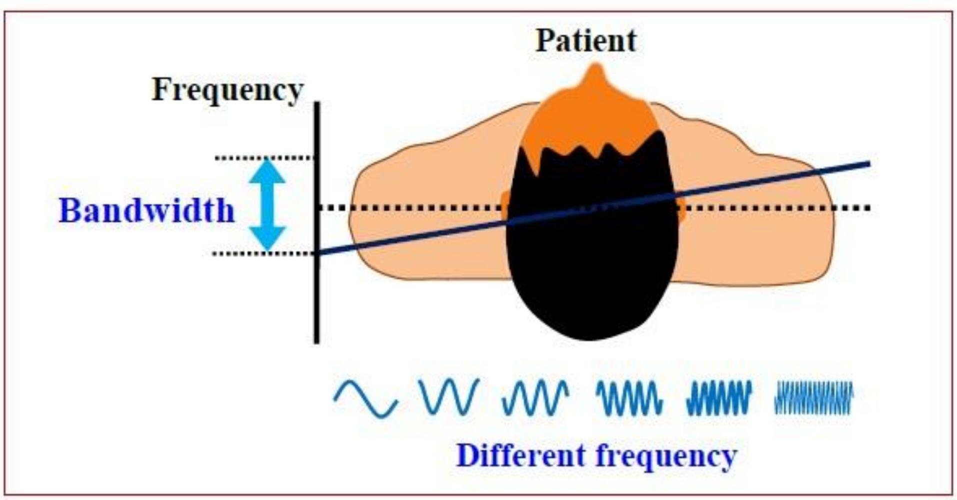

Figure 35: The different frequency of protons (**G_x** gradient: Frequency encoding gradient).

By causing this range of frequencies to exist, we can use the Fourier transform to separate them out after we measure an MRI signal as shown in Figure 36 (which is a mix of all signals from a slice).

They will accumulate an additional phase shift because of the different frequency, but – and this is utterly important - the already acquired phase difference, generated by the Phase Encoding gradient in the previous step, will remain. Now it is possible to determine whether the signal comes from the left, center or right-hand side of the slice. We can pinpoint the exact origin of the signals, which are received by the coil.

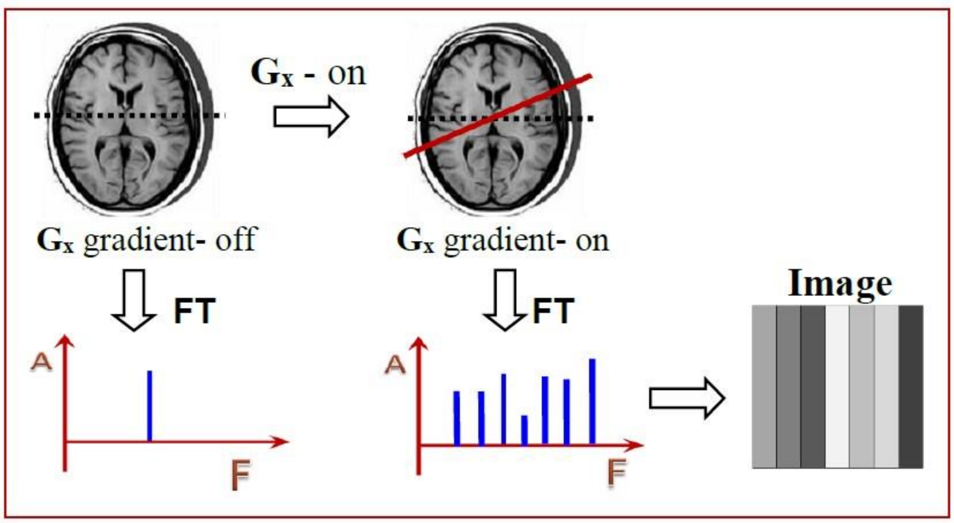

Figure 36: Fourier transform to separate frequency

Application of the Frequency Encoding Gradient

The frequency encoding gradient applied during the recording of the MRI signal. The frequency encoding gradient causes the precession of net magnetizations within the slice to be position dependent in that direction, whilst the signal is being recorded (see figure 37).

If the frequency encoding gradient is not applied at the same time as measuring the MRI signal, the signals from the different columns in the imaging slice will all have the same frequency. We cannot, therefore, use the Fourier transform to separate them out.

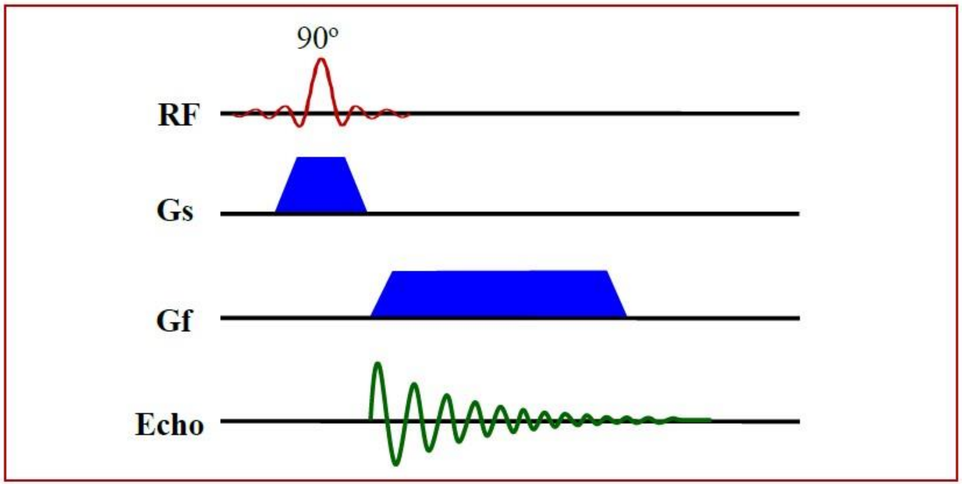

Figure 37: The frequency encoding gradient (Gf) is applied during signal measurement.

1.16.3 Phase Encoding Gradient

In order to encode the image or the imaged object along the so-call phase encoding direction, here the y direction (but the direction could also be in z or x direction), a gradient along the y direction is applied, and thereafter the signal is sampled. (G_{ph}) The phase encoding grading is a magnetic field gradient that allows the encoding of the spatial signal location along a second dimension by different spin phases. The phase encoding gradient is applied after slice selection and excitation (before the frequency encoding gradient), orthogonally to the other two gradients. The spatial resolution is directly related to the number of phase encoding steps (gradients). In fact, it is necessary to apply this gradient several times, each time increasing the gradient by an equidistant amount.

This is done by a gradient field is briefly switched on and then off again at the beginning of the pulse sequence right after the radio frequency pulse, the magnetization of the external voxels will either precess faster or slower relative to those of the central voxels.

During readout of the signal, the phase of the x-y-magnetization vector in different columns will thus systematically differ. When the x- or y- component of the signal is plotted as a function of the phase encoding step number *n* and thus of time *n* TR, it varies sinusoidally, fast at the left and right edges and slow at the center of the image. Voxels at the image edges along the phase encoding direction are thus characterized by a higher 'frequency' of rotation of their magnetization vectors than those towards the center.

As each signal component has experienced a different phase encoding gradient pulse, its exact spatial reconstruction can be specifically and precisely located by the Fourier transformation analysis. Spatial resolution is directly related to the number of phase encoding levels (gradients) used. The phase encoding direction can be chosen, e.g., whenever oblique MR images are acquired or when exchanging frequency and phase encoding directions to control wrap around artifacts.

In an MRI sequence diagram this procedure is indicated by the phase encoding (see Figure 38). As seen this figure consists of 32 steps in this example, each lasting for 0.250 ms, and with an increment of 0.734085 mT/m (milli Tesla pr meter). One could go through this figure from the bottom to the top, or from the top to the bottom, this is called linear phase encoding. One could also go from the middle and outward like 0, 1, -1, 2, -2 etc., this is called low-high phase encoding.

- Note that one particulate step corresponds to applying no gradient at all, and for the low-high encoding, this would then be the first step. In order to code the protons further the G_y gradient is switched on very briefly. During the time the gradient is switched on an additional gradient magnetic field is created in the Anterior-Posterior direction.

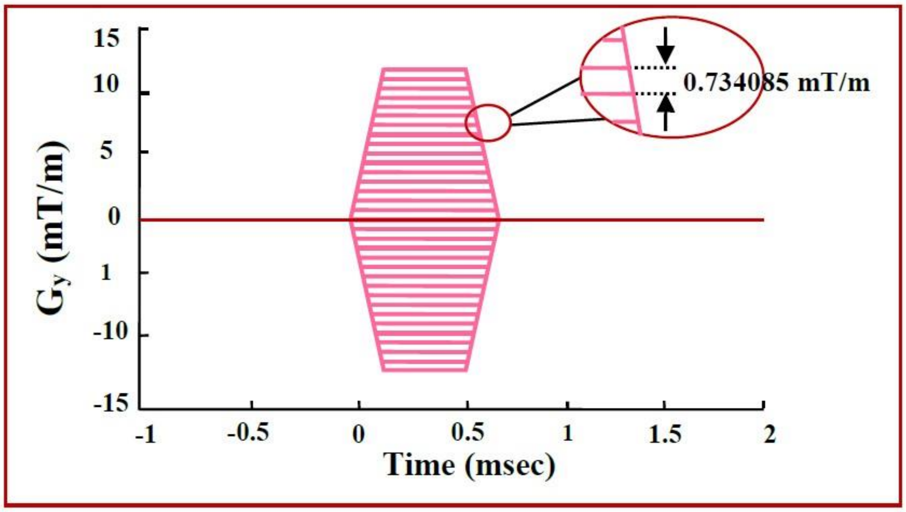

Figure 38: Phase encoding table - symmetrical

As you can see small volumes (voxels) have been created. Each voxel has a unique combination of frequency and phase. The number of protons in each voxel determines how strong (amplitude). The signal received contains a complex mix of frequencies, phases and amplitudes each from a different location (voxel) within the brain.

The computer receives this massive amount of information and then a "Miracle" occurs. In about 0.25 seconds the computer can analyze all this and create an image. The "Miracle" is a mathematical process, known as Two-Dimensional Fourier Transform (2DFT), which enables the computer to calculate the exact location and intensity (brightness) of each voxel.

This can be summarized (slice selection, phase encoding and frequency encoding) as follows:

After slice selection the encoding of spatial information has only to be performed in two dimensions. This can be accomplished by magnetic field gradients in the respective directions. These are differentiated by the time of gradient switching, i.e., before or during data acquisition. The first case, the so-called phase encoding, is discussed below. In the second case (frequency encoding), a readout gradient G_{RO} is switched during data acquisition and the gradient direction is therefore called the 'readout direction'. G_{RO} produces an additional, linearly varying magnetic field and due to the proportionality between magnetic field and frequency, the latter also alters linearly. Spins at different positions therefore emit radiation with different frequencies which can be distinguished after Fourier transformation. Each frequency is related to a specific position on the readout axis and the intensity of the radiation with this frequency is proportional to the number of spins emitting at this position.

الجامعة التقنية الوسطى
كلية التقنيات الصحية والطبية / بغداد
قسم : تقنيات الاشعة المادة: الفيزياء الشعاعية2 المرحلة: الثالثة

Title: **العنوان:**

GRADIENT SPECIFICATIONS

Name of the instructor: **اسم المحاضر:**

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Target population: **الفئة المستهدفة:**

3rd year طلبة المرحلة الثالثة

قسم تقنيات الاشعة

Posttest: **الاختبار البعدي:**

References: **المصادر:**

Thayalan, K., and Ramamoorthy Ravichandran. *The physics of radiology and imaging*. JP Medical Ltd, 2014.

1.17 Gradient Specifications

When you are shopping for an MRI scanner, it is very important to pay special attention to the gradient sub-system. Ideally, when a gradient is switched on it immediately reaches maximum power and when you switch it off the power is immediately back to zero (Figure 39A).

Unfortunately, this is not the case, as we do not live in an ideal world. In reality the gradient needs a little time to reach maximum power and to power down (Figure 39B). The time it takes to reach maximum power is called: Rise Time (Figure 39C). When we divide the maximum power by the rise time, we get a number called: Slew Rate. These are the specifications for a gradient system.

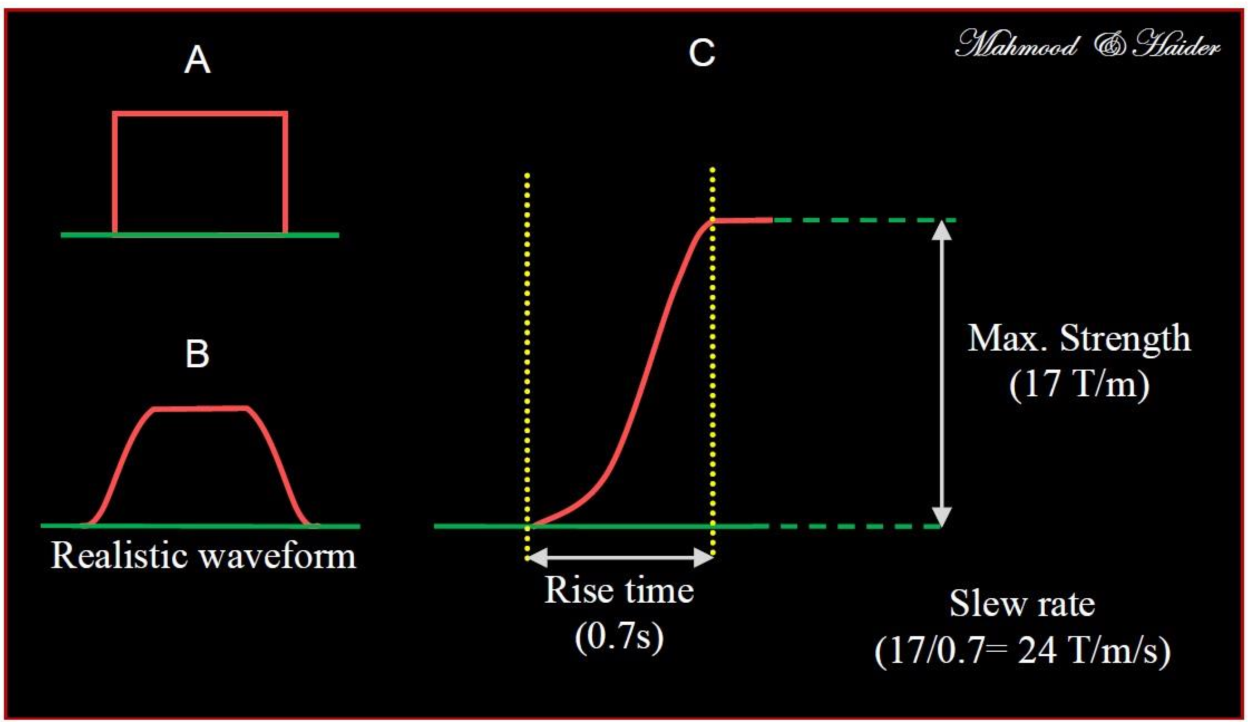

Figure 9.47: When a gradient is switched on it immediately reaches maximum power.

You should compare these values because they are different for each system:

1. Maximum strength: as high as possible (minimum FOV and maximum Matrix).
2. Rise time: as short as possible.
3. Slew rate: as big as possible (min. TR, TE and ETS).

The performance and therefore the range of applications, which can be done is mainly determined by the performance of the gradient system. Other issues you may look for are the field strength B_o, the computer system and the ease of use of the user interface.

1.18 MRI Image Quality, Artifacts, and Imaging Parameters

This part covers the factors that affect the quality of an MR image quality which depends on several factors:

- **Signal-to-noise ratio** (SNR) is equal to the ratio of MR signal received from the tissue being imaged to the background noise.

- **Contrast resolution** looks at the different and subtle differences in signal intensities from tissues being imaged.

- **The spatial resolution** of the image is the visualization of detail in the MR image. Therefore, it could be saying, the spatial resolution is the ability to distinguish between two points as separate and distinct. It is controlled by the voxel size. Spatial resolution may be increased by selecting: thin slices, Fine matrices and small FOV.

Sequence parameters such as Repetition time (TR), TE, slice thickness, field of view, and matrix size can affect these image quality issues adversely. Conversely, such parameters can have a positive effect on the quality of the image. Artifacts in MR are also a big issue with regards to image quality. Artifacts can be caused by a variety of things, from the equipment to the patient. These different artifacts can be assessed and resolutions found to correct for them.

1.18.1 Signal to Noise and Contrast Resolution

We can be likened to the noise, such as interferences which present as an irregular granular pattern. The noise can be degrading image information (MR signal). Image noise results from a number of different factors but it comes mainly from the tissue of the patient's body (RF emission due to thermal motion) and electronics inherent in the imaging process. These factors can be classified into two classes first that are beyond the operator's control (the MR scanner specifications and pulse sequence design) and on factors that the user can change:

1. **Fixed factors:** Imperfections of the MR system such as magnetic field inhomogeneities, thermal noise from the RF coils, pulse sequence design, patient-related factors resulting from body movement or respiratory motion.

2. **Factors under the operator's control**

- RF coil to be used
- Sequence parameters: voxel size (limiting spatial resolution), number of averaging, receiver bandwidth

The relationship between the MR signal and the amount of image noise present is expressed as the signal-to-noise ratio (SNR). The signal is the voltage induced in the receiver coil by the net magnetization when moved into the transverse plane. In other words, the signal comes from the excited protons on the selected slice plane. The SNR has a direct effect on the contrast resolution. The definition of contrast resolution is the difference in SNR between two adjacent areas. If the SNR is improved, then the contrast resolution of the image is improved; if the SNR is low, then the contrast resolution of the image is poor. Mathematically, SNR can be expressed as the intensity of the signal measured in the region of interest divided by the standard deviation of the signal intensity in a region outside the anatomy or the object being imaged (i.e., a region from which no tissue signal is obtained). The SNR is dependent on the following parameters:

1. Slice thickness and receiver bandwidth
2. Field of view
3. Size of the (image) matrix
4. Number of acquisitions
5. Scan parameters (TR, TE, flip angle)
6. Magnetic field strength
7. Selection of the transmit and receive coil (RF coil)

Repetition time (TR) is the interval between two successive excitations of the same slice. That means, it is the length of the relaxation period between two excitation pulses and is therefore crucial for T_1 contrast.

1.18.2 Pixel, Voxel, Matrix

Images that we get from MRI are digital images consist of a matrix of pixels (picture elements). Knowing that, the matrix is mathematically well-known two-dimensional grid of rows and columns. Each square of the grid is a pixel, which assigns the value corresponding to the signal intensity. Each pixel of the MR image corresponding three-dimensional volume element called voxel therefore provides information on the pixel corresponding voxel, (Figure 40).

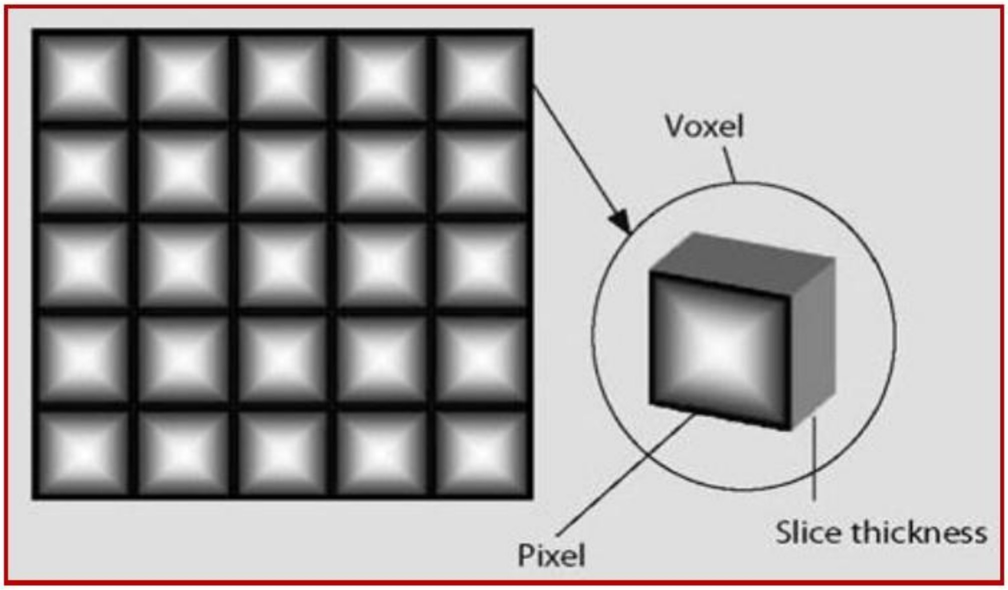

Figure 40: A voxel is the tissue volume represented by a pixel in the two-dimensional MR image.

Title: **العنوان:**

INTER-SLICE GAP

Name of the instructor: **اسم المحاضر:**

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Target population: **الفئة المستهدفة:**

3rd year طلبة المرحلة الثالثة

قسم تقنيات الاشعة

Posttest: **الاختبار البعدي:**

References: **المصادر:**

Thayalan, K., and Ramamoorthy Ravichandran. *The physics of radiology and imaging*. JP Medical Ltd, 2014.

1.18.3 Inter-Slice Gap

Spacing or inter-slice gap is the small space between two adjacent slices and can be measured by millimeters. It provides method to compensate for imperfect RF excitation pulse. Inter-slice gap allows the technologist to control the size of the imaging volume by increasing and decreasing space in between slices. Because the resultant slice profiles are not perfectly rectangular (Figure 41), two adjacent slices overlap at their edges when closely spaced. Therefore, it would be desirable to acquire contiguous slices but inter-slice gaps are necessary in spin echo (SE) imaging.

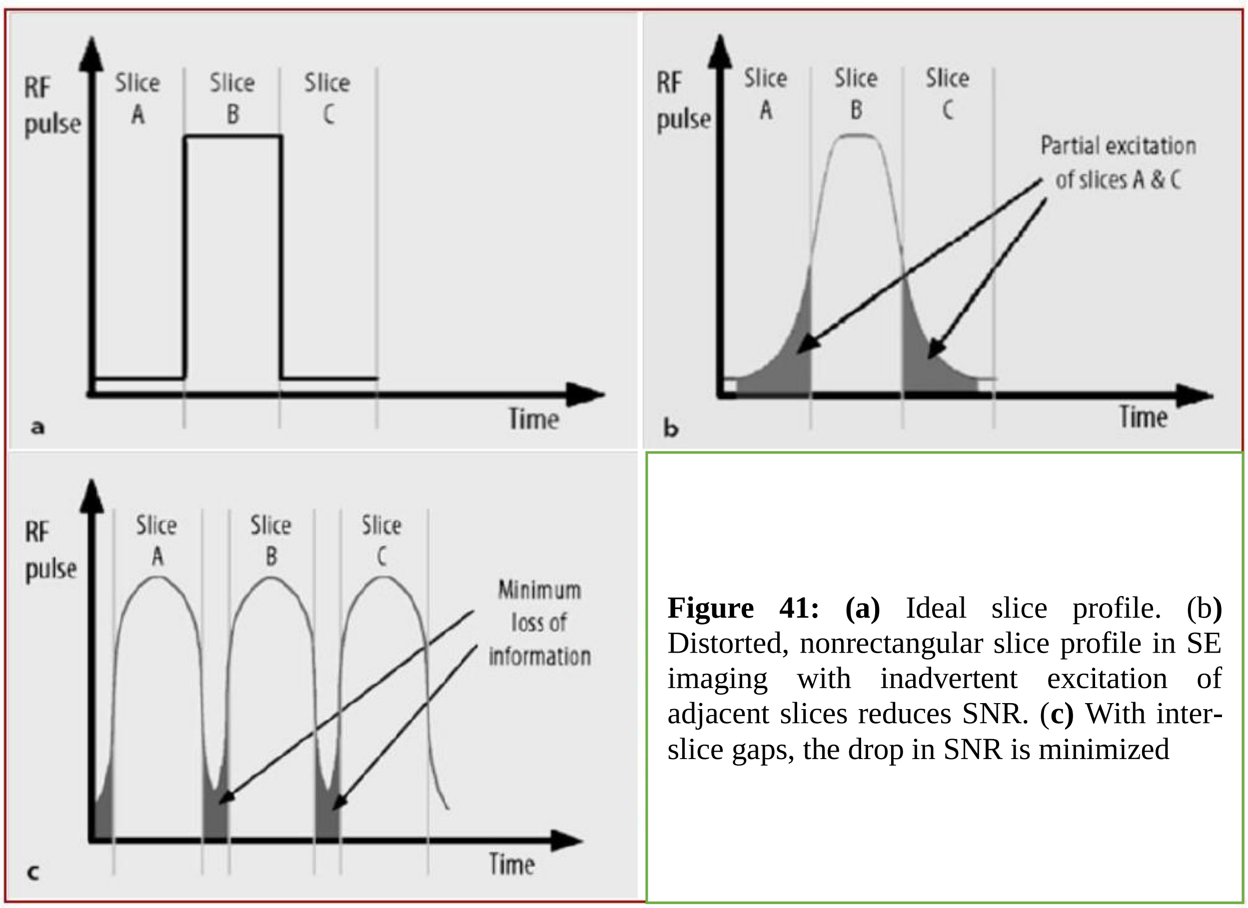

Figure 41: (a) Ideal slice profile. (b**)** Distorted, nonrectangular slice profile in SE imaging with inadvertent excitation of adjacent slices reduces SNR. (**c)** With inter-slice gaps, the drop in SNR is minimized

The RF pulse for one slice also excites protons in adjacent slices. Such interference is known as cross-talk. That is, when radio frequency pulse for one slice stimulates protons in the adjacent slices. The cross-talk will lead to a reduction SNR (Figure 41b). Therefore, when insert small gaps with thirty percent of slice thickness in between slices to minimize the artifact and improve signal to noise ratio. Because the resultant slices profiles are not perfectly rectangular (Figure 41). In selecting an appropriate inter-slice gap, one has to find a compromise between an optimal SNR, which requires a large enough gap to completely eliminate cross-talk, and the desire to reduce the amount of information that is missed when the inter-slice gap is too large. In most practical applications insert small inter-slice gaps with 25–50% of the slice thickness noise ratio. Alternative way is reducing the saturation of protons in adjacent slices a situation which the undesired by-slice imaging.

1.18.4 Size of the (image) Matrix

Another factor affecting signal to noise and contrast resolution is the voxel volume (3-dimensional volume of tissue) which is represented on the image matrix by a pixel (or picture elements). Spatial resolution corresponds to the size of the smallest detectable detail. The smaller the voxels are, the higher the potential spatial resolution will be.

Three parameters affect the Voxel volume **(**size of the voxel):

- **Pixel size**, which is established when the matrix size is chosen (256 × 256 or 512 × 512 etc...)
- **Field Of View (FOV)** (area of interest) (10 cm, 20 cm, etc.... the small FOV is usually less than 18 cm and the large FOV is more than 30 cm). The field of view (FOV) defines the image size in two (for 2D scans) or three (for 3D scans) dimensions and
- **Slice thickness**.

Matrices are two types:

- **Coarse matrices**: they have a small number of pixels in the field of view. and

- **Fine matrices**: they have a large number of pixels in the field of view.

An example of a coarse matrix is 128 × 128, whereas a fine matrix is 512 × 512 (see figure 42).

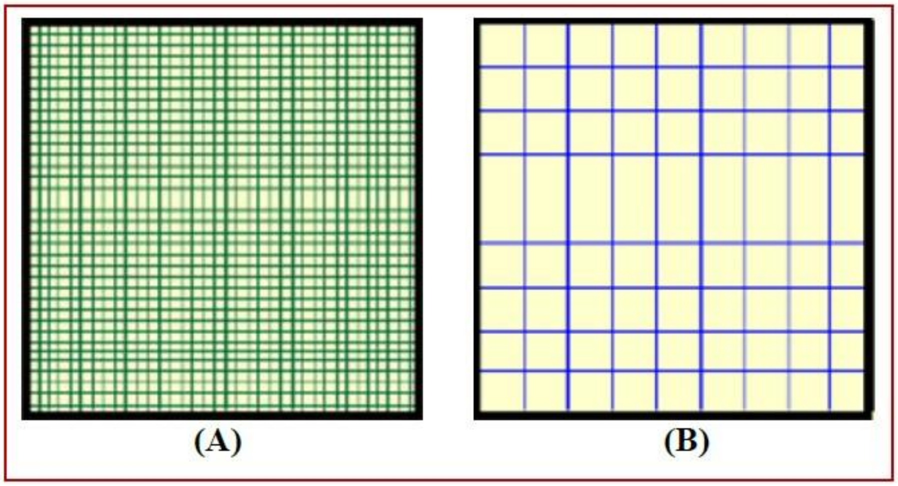

Figure 42: Types of matrix: (A) Fine Matrix (B) Coarse Matrix.

Voxel size determines the spatial resolution of the MR image. The size of a voxel can be calculated from the field of view, the matrix size, and the slice thickness. In general, the resolution of an MR image increases as the voxel size decreases.

Larger voxels have an increased signal to noise and better contrast resolution because there are more hydrogen nuclei in the voxel to contribute to the signal. Larger voxels are therefore represented on the image matrix by larger pixels.

Matrix size chosen establishes a pixel size and therefore the size of the voxel it represents. Another way to alter voxel size is with the slice thickness used. Assuming the field of view is square, doubling the slice thickness of the area doubles signal to noise ratio and voxel volume; thereby increasing contrast resolution. Halving the slice thickness adversely affects the signal to noise ratio and therefore decreasing the contrast resolution by half. The field of view also can influence the voxel volume. Doubling the field of view doubles, the voxel volume on both axes and increases the signal-to-noise by four. This also increases the contrast resolution of the image.

This is the single best and most efficient way to increase signal to noise ratio and contrast resolution. Halving the field of view reduces the voxel volume and reduces the signal to noise ratio by a quarter. The contrast resolution is decreased.

[The background noise that comes from the system is a constant amount for each patient, but is different for every patient namely that impact is different from one patient to another. Factors affecting the signal amplitude from the tissue affect the noise. The best pulse sequence for the signal amplitude is the classic spin echo (SE) sequence. Its use of the 180-degrees radio frequency pulse to re-phase all of the hydrogen protons in order to create an echo allows for the best signal amplitude. Other sequences such as the variations of gradient echo do not re-phase the hydrogen nuclei as effectively and signal is lost. The number of hydrogen protons in the area of tissue to be scanned has an effect on SNR and contrast resolution. If there are a large number of hydrogen protons in the area, then the signal amplitude will be increased; therefore, the contrast resolution will be increased. If the number of protons in the area is low, then the signal will be low and the contrast resolution will be poor.]

Title: **العنوان:**

SCAN PARAMETERS (TR, TE, FLIP ANGLE)

Name of the instructor: **اسم المحاضر:**

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Target population: **الفئة المستهدفة:**

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Posttest: **الاختبار البعدي:**

References: **المصادر:**

Thayalan, K., and Ramamoorthy Ravichandran. *The physics of radiology and imaging*. JP Medical Ltd, 2014.

1.18.5 Scan Parameters (TR, TE, Flip Angle)

Two controls determine tissue contrast: TR (repetition time) and TE (echo time) of the scan. They can be used for example to produce contrast between different tissues due to their individual relaxation properties. TR and TE both affect signal-to-noise and contrast resolution.

(a) Repetition time (TR) is the time between successive RF pulses, that is, the duration of a phase encoding cycle. A long TR allows the protons in all of the tissues to relax back into alignment with the main magnetic field. A short repetition time will result in the protons from some tissues not having fully relaxed back into alignment before the next measurement is made decreasing the signal from this tissue. In other words, TR controls the T1 relaxation time of the tissue by allowing a certain amount of the net magnetization to re-grow into the longitudinal plane, back to equilibrium before a signal is read. A long TR will increase signal to noise ratio because more net magnetization has re-grown back to equilibrium and is available to be excited and flipped once again into the transverse plane. A short TR decreases the signal to noise ratio because not as much of the net magnetization has recovered and is not there to be excited and flipped again into the transverse plane.

(b) Echo time (TE) is the time at which the electrical signal induced by the spinning protons is measured. That is, the time between giving the RF pulse (excitation) and the peak (maximum amplitude) of the echo signal (Fig 43). During this time interval, the transverse magnetization decays, e.g., signal decays, due to the T2 relaxation effects. So, TE directly determines how much the transverse signal decays. For a T2 weighted image, use a TE that is longer than the T2 of some tissues but shorter than the T2 of other tissues.

A long TE results in reduced signal in tissues like white matter and gray matter since the protons are more likely to become out of phase. Protons in a fluid will remain in

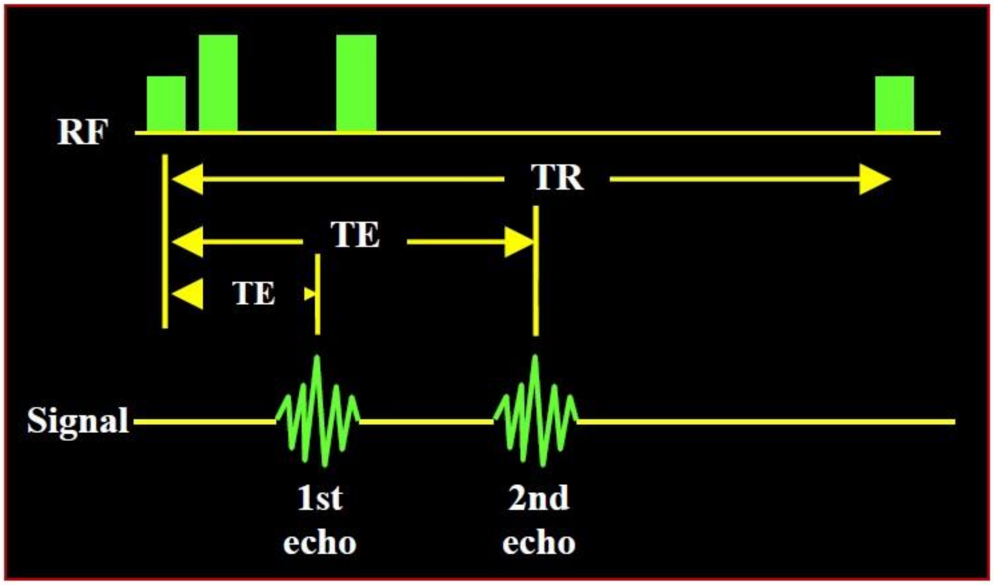

Figure 43: Echo time (TE) and Repetition time (TR).

phase for a longer time since they are not constrained by structures such as axons and neurons. A short echo time reduces the amount of dephasing that can occur in tissue like white matter and gray matter. In other words, TE controls the T2 relaxation time of the tissue by allowing a certain amount of the net magnetization to decay in the transverse plane before a signal is read. A long TE decreases signal to noise because all of the net magnetization has decayed when the signal is read. A short TE increases signal-to-noise because there is net magnetization in the transverse plane to contribute to the signal.

A long TR and a short TE increase signal to noise ratio and contrast resolution of the MR images. A specific weighting for the combination short TR and a long TE decrease signal-to-noise ratio and contrast resolution in MR. The results three parameters discussed so far are can be summarized in the Figure 44 together with several other important variables.

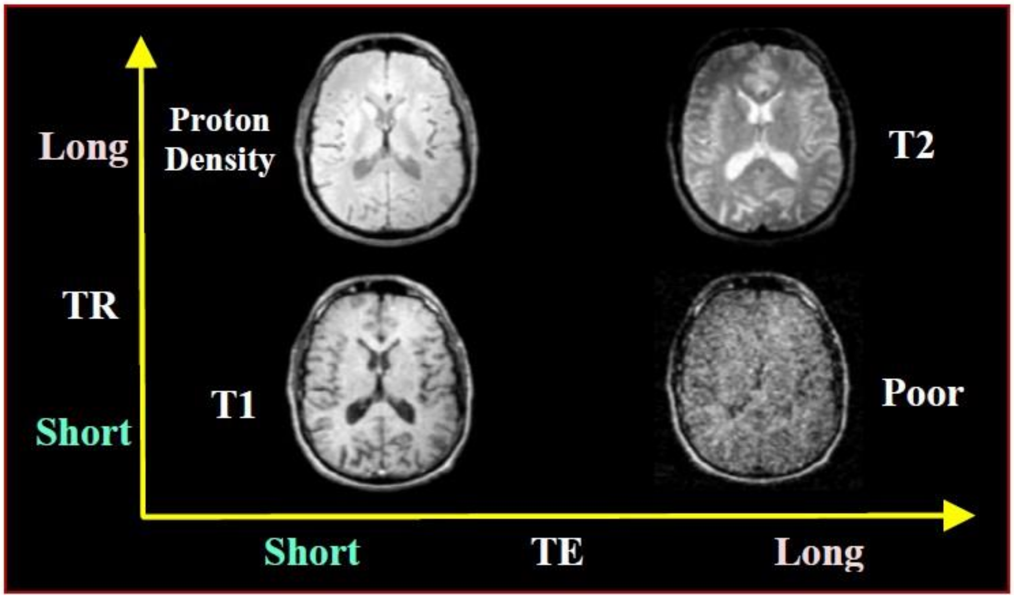

Figure 44: Summarize of TR-TE combinations which also shows some brain images impressively demonstrating the effects of relaxation weighting on the contrast.

1.18.6 Number of Acquisitions

The NEX, averages, or acquisitions represent how many times the tissue is sampled per TR. The number of excitations (NEX) or number of signal averages (NSA) denotes how many times a signal from a given slice is measured. The averages control the amount of data per line of K-space. If the averages are doubled, the data is doubled, so an increase occurs in the signal to noise ratio. Increasing the averages when using a low field strength magnet is necessary to maintain signal to noise ratio. The SNR, which is proportional to the square root of the NEX, improves as the NEX increases, but scan time also increases linearly with the NEX. Doubling the NEX or averages increases the signal to noise ratio (NSR) by the square root of 2 or by 1.44 times. Manipulating the NEX or averages is therefore not the best way to increase the signal to noise ratio and the contrast resolution. Increasing the NEX also increases the scan time and allow motion artifacts to appear on the images.

A parameter that is not often manipulated to increase image quality is the receive bandwidth. The receive bandwidth is the range of frequencies sampled by and during the readout gradient application. If the receive bandwidth is decreased, the signal to noise ratio is increased because less noise is picked up inherently by the readout gradient. If the receive bandwidth is cut by a half, the signal to noise ratio is increased by forty percent. However, the sampling time, or the time the readout gradient is left on, must be increased. A decreased bandwidth also could increase the minimum TE that can be chosen because of the increased time the readout gradient is left on. The readout or frequency encoding gradient is turned on usually during the re-phasing, peak, and de-phasing cycle of the echo or signal. If the readout or frequency gradient must be left on longer, then a short, short TE cannot be used because all of these pieces of the cycle have to occur. A decreased receive bandwidth increases signal to noise and also contrast resolution.

1.18.7 Field of View

There is a close relationship between field of view (FOV) and SNR. When matrix size is held constant, the FOV determines the size of the pixels. Pixel size in the **frequency-encoding** direction is calculated as:

FOV in mm divided by the matrix in the frequency-encoding direction and

Pixel size in the **phase-encoding** direction is calculated as:

FOV in mm divided by the matrix in the phase-encoding direction.

Another limiting factor is image acquisition or scan time, which increases in direct proportion to the matrix size. Scan time is the key to the economic efficiency of all MR systems and can be calculated by a simple equation.

Scan time = TR × number of phase-encoding steps × number of signal averages (NSA) [echo train length (ETL)].

1.18.8 Selection of the Transmit and Receive Coil (RF Coil)

Surface coils are pads or pieces of equipment that are placed next to a body part in order to enhance signal from the tissue. The correct size of the coil must match the size of the body part to be imaged in order for optimal image quality. Quadrature coils have a better signal to noise than most because it is made up of two receiver coils to get the signal with. Surface coils in general increase signal to noise as well as contrast resolution if used properly.

The most beneficial ways to increase SNR and contrast resolution in MR images is the use of spin echo imaging, long TR and short TE, coarse matrices with a large field of view, thick slices, and increased NEX or acquisitions.

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MRI CONTRAST AGENTS

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References: **المصادر:**

Thayalan, K., and Ramamoorthy Ravichandran. *The physics of radiology and imaging*. JP Medical Ltd, 2014.

1.19 MRI Contrast Agents

'Contrast' refers to the signal differences between adjacent regions, which could be 'tissue and tissue', 'tissue and vessel', and 'tissue and bone'. The inherent difference in T1 relaxation time between biological tissues, or between normal and pathologic tissue is not always large enough to obtain a detectable contrast in the MR image (cf. 45a). In order for pathology (or any process for that matter) to be visible in MRI, there must be contrast or a difference in signal intensity between it and the adjacent tissue. The contrast mechanism for MRI is more complicated and not as in the Contrast agents for X-ray and CT where show contrasting effects according to the electron-density difference, and they produce direct contrast effects on their positions.

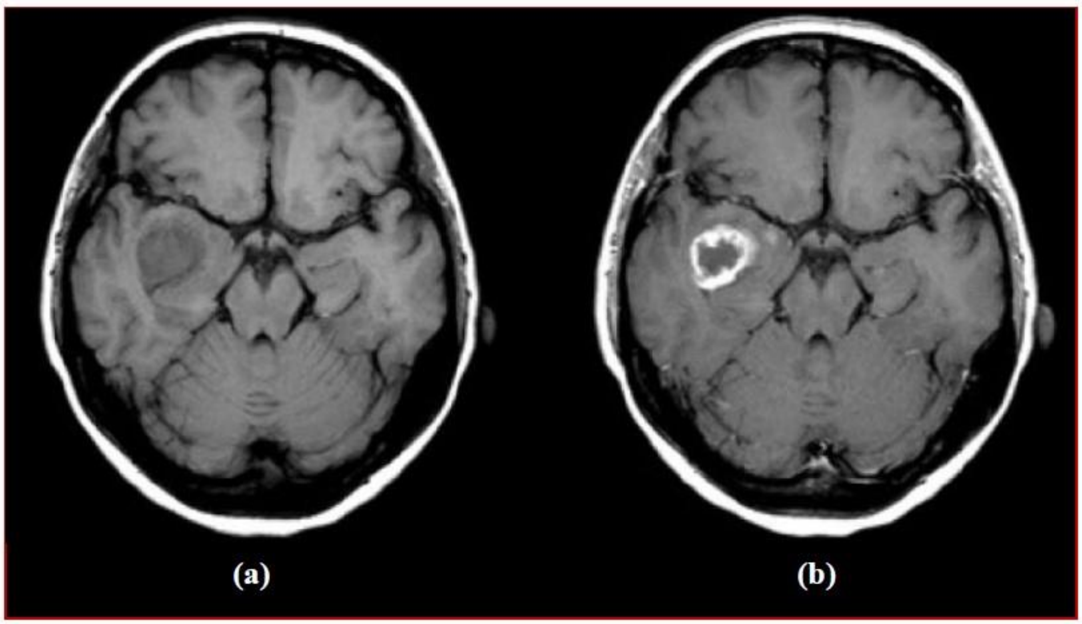

Figure 45: MR image to diagnose tumor in the brain without contrast agent (a) and with contrast agent.

For MRI, the contrast enhancement occurs as a result of the interaction between the contrast agents and neighboring water protons, which can be affected by many intrinsic and extrinsic factors such as proton density and MRI pulse sequences. As we know from previous subjects, the signal contrasts can arise in MRI from differences in four basic physical parameters:

SD: The spin density of the various tissues/fluids being analyzed

T1: The time constant with which the spin magnetization of a given tissue will build up after being saturated/inverted/pulsed-away

T2*: The time constant with which the spins' signals arising from a given tissue will diphase due to inhomogeneous broadening –this is the kind of signal decay that can be echoed away by π pulses; for instance, the one arising from field inhomogeneities or susceptibility differences

T2: The (longer) time constant with which the spins' signals arising from a given tissue will decay away due to homogeneous broadening – this is the kind of irreversible decay that can't be echoed away, arising from microscopic random fluctuations in the magnetic field.

Contrast in most MR images is actually a mixture of all these effects; but careful design of the imaging pulse sequence allows one contrast mechanism to be emphasized while the others are minimized. The ability to choose different contrast mechanisms by tailoring the appropriate pulse sequence and choosing the right pulse sequence parameters is what gives MRI its tremendous flexibility.

In certain cases, the intrinsic differences in T1, T2, T2*, etc., may not be sufficient to achieve the desired degree or kind of contrast. In those cases, additional differences can be introduced by adding **contrast agents** (see figure 45b): paramagnetic chemicals that localize in certain tissues/fluids, and artificially change their spin relaxation properties.

In order for an excited spin system to return to its equilibrium magnetization, energy must be transferred from the spin system to the lattice (surrounding), as discussed in section 9.13. The return to equilibrium is described by the spin-lattice relaxation time, T1. When T1-weighted sequences are used, the magnitude of the MR-signal increases with decreasing T1-relaxation times. Further, the contrast between two tissues will of course also increase with increasing difference in T1 relaxation times between the two tissues. Sufficient contrast is of particular importance in differentiating pathological tissue from normal surrounding tissue. Exogenous MR contrast agents were therefore developed shortly after the first commercial MR systems became available in the early 1980's. Today, MR contrast agents are typically in a significant proportion of MR examinations; with the highest usage in CNS applications (tumor diagnosis). MR contrast agents are also widely used in MR angiography (MRA). MR contrast agents act by selectively reducing T1 (and T2) relaxation times of tissue water through spin- interaction between electron spins of the metal-containing contrast agent and water protons in tissue

There are two classes of MRI contrast agents available,

(1) T_1-weighted contrast agents (e.g., gadolinium-(Gd^{3+}) and manganese- (Mn^{2+}) chelates) are paramagnetic in nature which increase the T_1 relaxation time, resulting in bright contrast T_1-weighted images; and

(2) T_2-weighted contrast agents are superparamagnetic materials (e.g., magnetite (Fe_3O_4) nanoparticles) which reduce T_2 relaxation times, giving rise to dark contrast T_2-weighted images. The efficiency of a contrast agent to reduce the T_1 or T_2 of water protons is referred to as relaxivity and defined by followed equation:

$$\boldsymbol{1/T_{1,2} = 1/T_{01,2} + \gamma_{1,2}\mathrm{C}}$$

Where $1/T_{1,2}$ is the observed relaxation rate in the presence of contrast agents, $1/T_{01,2}$ is the relaxation rate of pure water, C is the concentration of the contrast agents and $r1$ and $r2$ are the longitudinal and transverse relaxivities, respectively

Various inorganic nanoparticles have been used as magnetic resonance imaging (MRI) contrast agents due to their unique properties, such as large surface area and efficient contrasting effect. Since the first use of superparamagnetic iron oxide (SPIO) as a liver contrast agent, nanoparticulate MRI contrast agents have attracted a lot of attention. Magnetic iron oxide nanoparticles have been extensively used as MRI contrast agents due to their ability to shorten T_2* relaxation times in the liver, spleen, and bone marrow. More recently, uniform ferrite nanoparticles with high crystallinity have been successfully employed as new T_2 MRI contrast agents with improved relaxation properties. Iron oxide nanoparticles functionalized with targeting agents have been used for targeted imaging via the site-specific accumulation of nanoparticles at the targets of interest. Recently, extensive research has been conducted to develop nanoparticle-based T_1 contrast agents to overcome the drawbacks of iron oxide nanoparticle-based negative T_2 contrast agents. In this report, we summarize the recent progress in inorganic nanoparticle-based MRI contrast agents.

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Posttest: الاختبار البعدي:

References: المصادر:

Thayalan, K., and Ramamoorthy Ravichandran. *The physics of radiology and imaging*. JP Medical Ltd, 2014.

1.20 K-Space

K-space is a formalism widely used in magnetic resonance imaging introduced in 1979 by Likes and in 1983 by Ljunggren and Twieg, which form raw data matrix in MRI which can be converted into an image using Fourier transformation (Figure. 46).

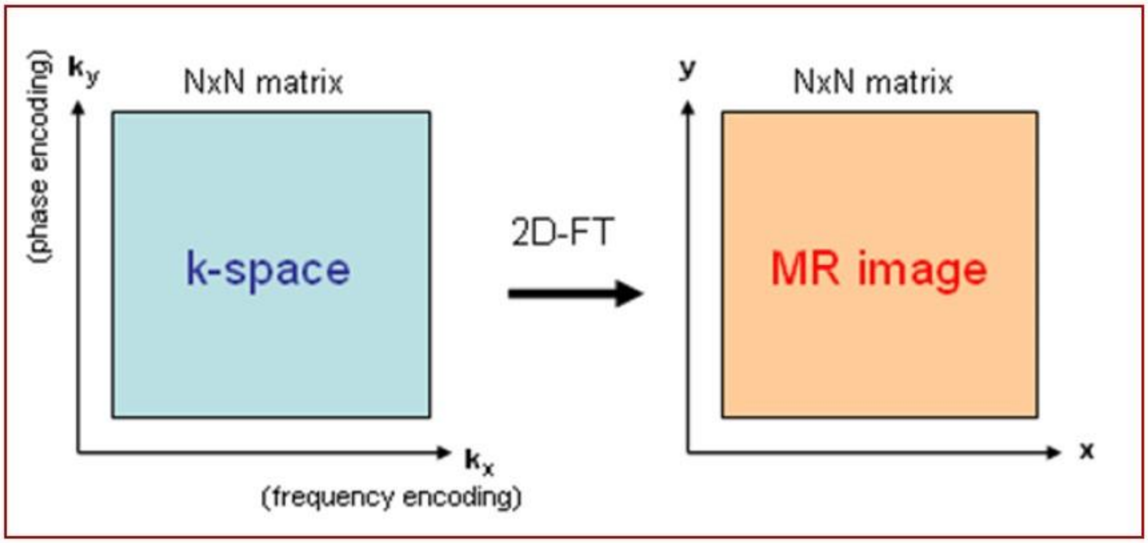

Figure 46: Raw data matrix in MRI converted into an image using FT.

The value k_x defines the number of phase cycles per meter distance from the origin ($x = 0$) a magnetization vector passes through due to application of the magnetic field gradient (G_x). By analogy to the frequency given as cycles per time, k is called the "spatial frequency". To illustrate the idea of k-space to start with the following question: Why is k-space so important?

The answer is: It helps us to understand how an MRI image is acquired and how various pulse-sequences work.

In MRI physics, k-space is the 2D or 3D Fourier transform of the MR image measured. Its complex values are sampled during an MR measurement, in a premeditated scheme controlled by a pulse sequence, i.e., an accurately timed sequence of radiofrequency and gradient pulses. In practice, k-space often refers to the temporary image space, usually a matrix, in which data from digitized MR signals are stored during data acquisition. When k-space is full (at the end of the scan) the data are mathematically processed to produce a final image. Thus k-space holds raw data before reconstruction.

“The MRI data prior to becoming an image (raw or unprocessed data) is what makes up k-space”. Synonyms for k-space are matrix and time time-domain. The task of an MRI scanner is to recognize and collect MR signals and store them in a specific order which is recognizable for further analysis. At each RF excitation, combinations of different excitations are collected as one complex signal. The read-out MR signal is stored in a 2D array called k-space, containing samples of the continuous Fourier transform of the object's magnetization.