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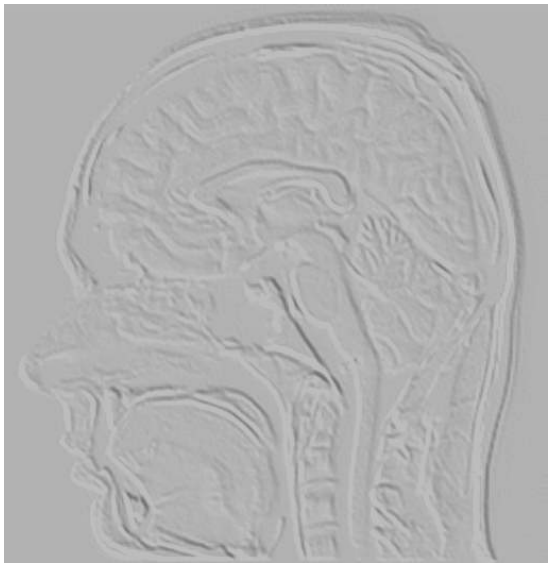

mri : Physics

For anyone who has not a degree in physics

Evert J Blink
Application Specialist MRI

Preface

Over the years Magnetic Resonance Imaging, hereafter referred to as MRI, has become a popular and widely available means of cross sectional imaging modality. That is not coincidental; MRI has gone through a fast paced round of development since its discovery. Now every self-respecting hospital or clinic has one or more MRI scanners to battle the conquest for more precise and accurate imaging and diagnosis of pathology. Even as we speak the development is still in full swing.
Paired with its excellent image contrast resolution, MRI is harmless to the human body, within reason, through its use of radio waves and a magnetic field. This in contrast with X-ray's and CT examinations, which use ionizing radiation.

As MRI becomes more and more accepted the need for more qualified staff is also increasing. Through the years operation of MRI scanners has become easier with each new software release, but this does not eliminate the need for proper understanding of how MRI works. MRI works with a host of parameters, such as TR, TE, Flip Angle, Phase Encoding to name but a few. A thorough understanding of these parameters is vitally important in order to produce a successful MR image.

There are numerous books on the bookshelves about MRI physics, most of them aimed at those amongst us who are already experienced and have a fair understanding of physics. Few books are written for the absolute beginner, who does not have a degree in physics.
As an application specialist I often have to explain the basic concept of MRI to people, most of the time radiographers, who do understand the physics related to X-ray, but never got into contact with the physics related to MRI. Having said that, nowadays radiography lectures also include MRI physics. Yet, these courses also use the same books aimed at experienced people.

What I am trying to do here is to write about MRI physics in such a way that everybody can understand the concept. It helps, of course, if one has been in contact with physics, but it is not absolutely necessary. Once you have a reasonable understanding of the concept you can go ahead and pick up one of the more advanced books.
One thing you should realize though. MRI physics is highly complex if you want to know it all. You can dig into quantum physics until you see green and still you may not be able to piece it all together. There are very few people who understand MRI to its full extent. The rest of us, mere mortals, grasp the basic concept. However, let this not discourage you, there is only so much you need to know, and luckily, that is not all that much, in order to do your job properly.

Let me take the liberty of offering you some advice: keep reading about MRI. Each time you reread the story you'll learn something new. And there will be a day that all the pieces come together.
When this happens you are invited to read the story again and you will discover that there is still more to learn.

Until then I hope this story will introduce you gently into the exciting world of MR Imaging: the one imaging modality that never gets boring.

Evert Blink
November, 2004

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A little MRI History

The story of MRI starts in about 1946 when **Felix Bloch** proposed in a Nobel Prize winning paper some rather new properties for the atomic nucleus. He stated that the nucleus behaves like a magnet. He realized that a charged particle, such as a proton, spinning around its own axis has a magnetic field, known as a magnetic momentum. He wrote down his finding in what we know as the Bloch Equations. It would take until the early 1950s before his theories could be verified experimentally. In 1960 Nuclear Magnetic Resonance spectrometers were introduced for analytical purposes. During the 1960s and 1970s NMR spectrometers were widely used in academic and industrial research. Spectrometry is used to analyze the molecular configuration of material based on its NMR spectrum.

In the late 1960s **Raymond Damadian** discovered that malignant tissue had different NMR parameters than normal tissue. He mused that, based on these differences, it should be possible to do tissue characterization. Based on this discovery he produced the first ever NMR image of a rat tumor in 1974. In 1977 Damadian and his team constructed the first super conducting NMR scanner (known as The Indomitable) and produced the first image of the human body, which took almost 5 hours to scan (*Figure 1*).

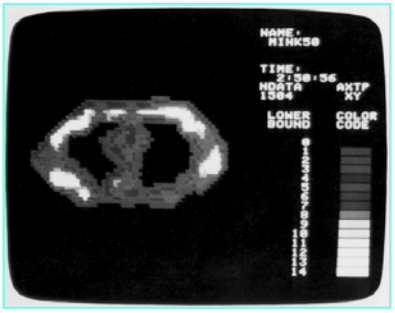


Figure 1

At the same time **Paul Lauterbur** was pioneering in the same field. One could discuss who was responsible for bringing MRI to us, although, in all fairness, one could accept that both gentlemen had their contribution.

The name Nuclear Magnetic Resonance (NMR) was changed into Magnetic Resonance Imaging (MRI) because it was believed that the word nuclear would not find wide acceptance amongst the public.

The rest is, as they say, history. In the early 1980s just about every major medical imaging equipment manufacturer researched and produced MRI scanners. Since then a lot has happened in terms of development. The hardware and software became faster, more intelligent and easier to use. Because of the development of advanced MRI pulse sequences more applications for MRI opened up, such as MR Angiography, Functional Imaging and Perfusion / Diffusion scanning.
And yet, the end is not in sight. The development of MR is still in full swing and only time will tell what the future has in store for us.

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 principle 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.

The Hardware

MRI scanners come in many varieties. It's like going to the supermarket; you're spoiled for choice. You can have permanent, resistive, superconducting, and open or bore type magnets, with or without helium, low or high field strength. What do you choose? The choice of magnet is mainly governed by what you intend to do with it and how much money you have in the bank. High field magnets offer better image quality, faster scanning and a wider range of applications, but they are more expensive than their low field counterparts.

Magnet Types

Permanent magnets

Figure 2

A permanent magnet consists of a material, which has been magnetized such that it won't loose its magnetic field, (like the ones you put on your refrigerator). The field strength is usually very low and ranges between 0.064T ~ 0.3T (the unit for magnetic field strength is Tesla. 1 Tesla = 10000 Gauss). Permanent magnets have usually an open design, which is more comfortable for the patient. *Figure 2* shows Toshiba's Access 0.064Tesla system. The Access was the worlds first open MRI scanner.

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

Resistive Magnets

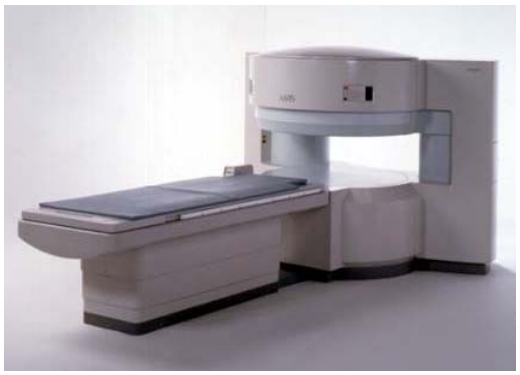


Figure 3

Resistive magnets are very large electro magnets, like the ones used in scrap yards to pick up cars. The magnetic field is generated by a current, which runs through loops of wire. Resistive magnets come in two flavours: air-core and iron-core. The field strength can be up to 0.3 Tesla. They produce a lot of heat, which requires water-cooling. They need a lot of power to run, and are usually switched off when not in use to conserve power. They usually have an open design, which reduces claustrophobia.
Figure 3 shows Hitachi's Airis 0.3 Tesla (air-core) system.

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

Superconducting magnets

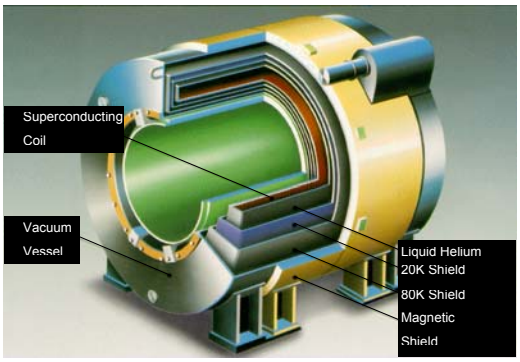


Figure 4

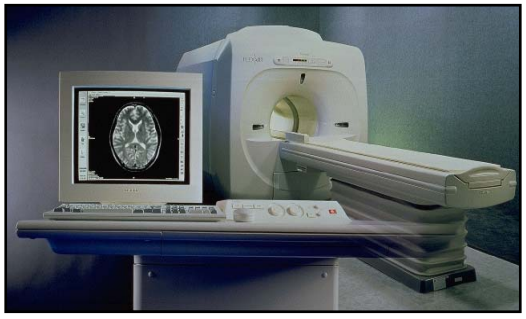


Figure 5

Today's most commonly used magnets are superconducting magnets. The magnetic field is generated by a current, which runs through a loop of wire. The wire is surrounded with a coolant, such as liquid helium, to reduce the electric resistance of the wire. At 4 Kelvin (-269º C) electric wire looses its resistance. Once a system is energized, it won't loose 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. *Figure 4* shows how a superconducting magnet is build up. 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.

Figure 5 shows a few examples of bore type magnets from various vendors.

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

Figure 6

In 1997 Toshiba introduced the worlds 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 6* shows Toshiba's OPART 0.35 Tesla system, which combines an open design with the advantages related to superconducting magnets.

Another advantage of open design magnets is the possibility to perform interventional procedures while scanning.

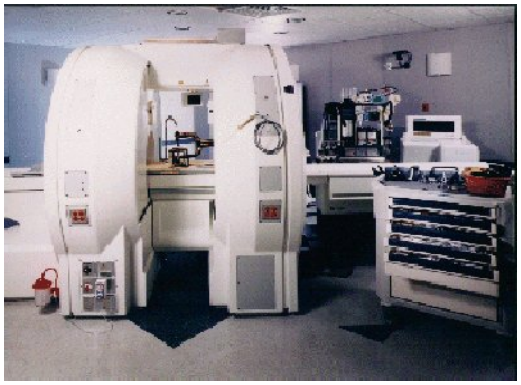


Figure 7

Figure 7 shows General Electric's superconducting "double donut" system operating at 0.5 Tesla. The surgeon stands between the two poles of the magnet. Although it offers the best design for interventional procedures when it comes to patient accessibility, the fact that two 1.5 Tesla systems have been used makes it extremely expensive.

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.

RF Coils

RF coils are needed to transmit and receive radio-frequency waves used in MRI scanners. RF coils are one of the most important components that affect image quality. Current MRI scanners have a range of RF coils suitable to acquire images of all body parts. There are two types of RF coils: volume coils and surface coils.

Volume RF Coils

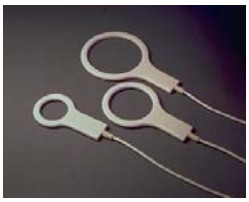
Head coil

Knee

Figure 8

The design of a volume coil is usually a saddle shape, which guarantees a uniform RF field inside the coil. 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 8 shows two volume coils. The head coil is a transmit/receive coil; the knee coil is receive only.

Surface coils



Surface coils

Shoulder coil

Neck coil

Spine coil

Figure 9

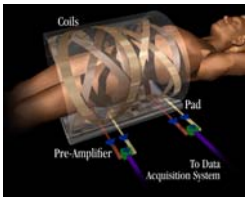
As the name already implies, surface coils are placed close to the area under examination such as the temporo-mandibular joint, the orbits or the shoulder. The coil consists of a single or double loop of copper wire. They have a high Signal to Noise Ratio (SNR) and allow for very high-resolution imaging. The disadvantage is that they loose 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.
Figure 9 shows a few examples of surface coils.

Quadrature Coils

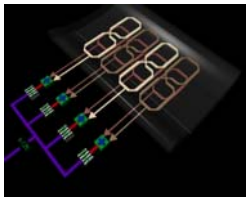
Quadrature or circularly polarized coils can have either a saddle shape or as a surface coil. What they have in common is that they contain at least two loops of wire, which are placed at right angles to one another. The advantage of this design is that they produce √2 more signal than single loop coils. Nowadays, most volume coils are Quadrature coils. The coils shown in *Figure 8* are Quadrature coils.

Phased Array Coils

Phased array coils consist of multiple surface coils. Surface coils have the highest SNR 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.



QD Body Array coil



Spine Array coil

Figure 10

Figure 10 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.

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. Actually, the receive coil is a part of the RF chain. 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".

Furthermore, one needs a processor to process the received signal, as well as to control the complex business of scanning.

Let's Talk Physics

Introduction

It is difficult to decide where to start when you want to explain the physics of MRI. You could say, "start at the beginning", and you're right, that's where all good stories start. But with MRI physics it is a bit more difficult, because one first has to establish where the beginning is or, put in another way, how much do you want to know.
As the title already indicates this story is supposed to entice those people who are new to the business and who need to know the very basics of MRI physics. In one way this is easy to write up because I can leave out large chunks of physics. On the other hand it is very difficult because I have to assume you know nothing and yet I have to explain something complicated in an easy to understand manner. Believe me, it's not at all that easy.
But let that be my problem. After all, it's my job ☺.

Magnetization

Let us start the journey into MRI physics by looking around us. What do we see? Amongst a host of items which have nothing to do with MRI we see the earth. There are a few things we know about earth:

1. The earth is a giant ball that floats in space. Not randomly, but that's a different story.
2. The earth has a moon, which rotates around the earth.
3. The earth has an electrical charge. Whether it be positive or negative is not important.
4. The earth rotates (spins) around it's own axis. This is the interesting bit.
5. There's a heck of a lot of water on earth, around 70% worth, and most of it seems to be falling in my back garden while I'm writing this.

This giant, 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 μT at the poles and 70 μT at the equator. (Tesla is for magnetic fields what Ampere is for electric current).

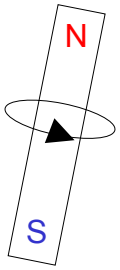


Figure 11

In short we can establish that the earth is a giant spinning bar magnet, with a north and a south pole (*Figure 11*). And don't forget it is wet, very wet.

You may wonder what all this has to do with MRI but I'll get to that in a moment.

Now let's have a look at ourselves, the Homo sapiens. What do we have in common with earth? Well on first sight not a lot, but when 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 moons floating around it. What we are looking at 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.

From our chemistry lessons we know that there are many different elements, 110 to be precise. Because we exist mainly of water let's have a look at it. Water consists of 2 Hydrogen and 1 Oxygen atom. The hydrogen atom (the first element in the periodic table) has a nucleus, called proton, and 1 moon, called electron.
This proton is electrically charged and it rotates around its axis. There we have the analogy with the earth. Also the hydrogen proton can be looked at as if it were a tiny bar magnet with a north and a south pole.

Why do we take hydrogen as our MR imaging source?
There are two reasons. 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 story 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.

For who really wants to know, Hydrogen is not the only element we can use for MR imaging. In fact any element, which has an odd number of particles in the nucleus, can be used. Some elements, which can be used, are:

Table 1: MRI friendly elements

Isotope	Symbol	Spin Quantum number	Gyro Magnetic Ratio (MHz/T)
Hydrogen	^{1}H	1/2	42.6
Carbon	^{13}C	1/2	10.7
Oxygen	^{17}O	5/2	5.8
Fluorine	^{19}F	1/2	40.0
Sodium	^{23}Na	3/2	11.3
Magnesium	^{25}Mg	5/2	2.6
Phosphorus	^{31}P	1/2	17.2
Sulphur	^{33}S	3/2	3.3
Iron	^{57}Fe	1/2	1.4

If we look at a bunch of hydrogen protons (as in a molecule) we see, in fact, a lot of tiny bar magnets spinning around their own axes. As we may recall from classes, two north poles and two south poles of two magnets repel each other, while two poles of opposite sign attract each other.
In our body these tiny bar magnets are ordered in such a way that the magnetic forces equalize. Our bodies are, magnetically speaking, in balance.
Just as well, otherwise we would attract a lot of metal when we go about.

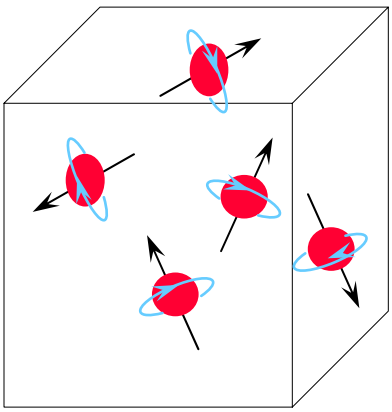


Figure 12

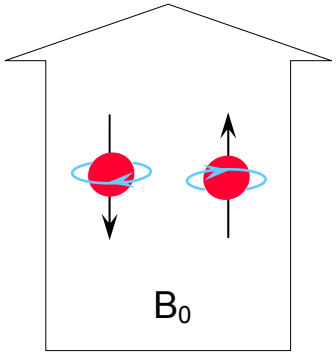
Now we have established some interesting facts of life, lets have a look at what happens when we go ahead and try to make an MRI examination.

As we have seen in the paragraph about the hardware, magnets used for MR imaging come in various field strengths. The magnetic field strength of a 1.5 Tesla magnet is ± 30.000 times stronger than the earth gravitational field! This indicates that we are working with

potentially dangerous equipment (more about that later).

When we put a person in a magnet 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.



(B_0 is the indication for the magnetic field of the MRI scanner)

Figure 13

2. They precess or "wobble" due to the magnetic momentum of the atom.

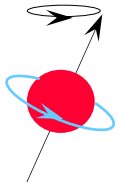


Figure 14

They precess at the Larmor frequency. This Larmor frequency is something, which needs a little further explanation. The Larmor frequency can be calculated from the following equation (don't worry, this will be the first of only two equations you will encounter in this story):

$$\omega_0 = \gamma B_0$$

Where: ω_0 = Precessional or Larmor frequency. (MHz)

γ = Gyro Magnetic Ratio. (MHz/T)

B_0 = Magnetic field strength. (T)

Here we see two things, which we discussed before, come together: the Gyro Magnetic Ratio and the Magnetic field strength.

So, there you have it. A cute little equation, if ever I saw one. But why is this so important? Well, we need the Larmor frequency to calculate the operating frequency of the MRI system.
If we have a MRI system of 1.5 Tesla then the Larmor or precessional frequency is: 42.57 x 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. You can view these values on your own system by checking the Centre Frequency or similar phrase.

Now we know what happens to the individual protons when we put a victim in the scanner. Lets continue the story and see what happens further.

When the protons experience a strong magnetic field from the scanner we saw that they can align with the field in two ways: parallel and anti-parallel. You could call this also Low and High Energy State.
The distribution of the protons for both states is not the same. The protons are, just like many people, lazy. They prefer to be in low energy state. There are more protons aligned parallel or low energy state than there are anti-parallel or high energy state (*Figure 15*). However, it's not that big a difference. The excess amount of protons aligned parallel within a 0.5T field is only 3 per million (3 ppm = parts per million), in a 1.0T system there are 6 per million and in a 1.5T system there are 9 per million. So, the number of excess protons is proportional with B_0. That is also the reason why 1.5T systems make better images than systems with lower field strengths.
9 ppm excess protons don't seem very many, but in real life it adds up to quite a number. Have a look at the following calculation made by **Moriel NessAiver**, Ph.D. (He wrote an excellent book about MRI physics, which I highly recommend. See § recommended reading). He 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 6.02×10^{23} molecules per mole.
- 1 mole of water weighs 18 grams ($O^{16} + 2H^{1}$), has 2 moles of Hydrogen and fills 18 ml, so……
- 1 voxel of water has $2 \times 6.02 \times 10^{23} \times 0.02 / 18 = 1.338 \times 10^{21}$ total protons
- The total number of excess protons =

$$\frac{1.338 \times 10^{21} \times 9}{2 \times 10^{6}} = \mathbf{6.02 \times 10^{15}} \text{ or } \mathbf{6 \text{ million billion}}!!!$$

Don't do this at home! (In other words: don't remember this)

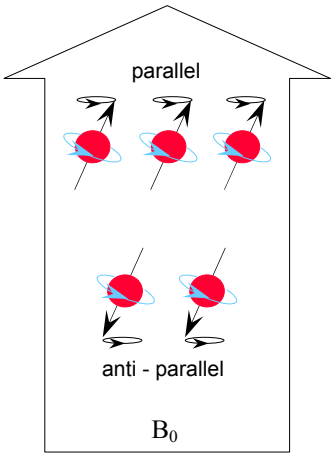


Figure 15

In the end we 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.
It is with this net magnetization that we continue.

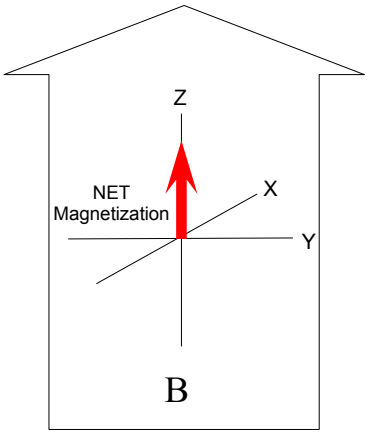


Figure 16

In order to see what happens with this net magnetization in our MRI experiment in an easy way, the scientific community came up with the brilliant idea to visualize it by means of vectors.
A vector (the red arrow in the *Figure 16*) has a direction and a force. To see what happens with the vector (net magnetization) we imagine a frame of rotation, which is nothing else than 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_z or longitudinal magnetization.
It is now possible to make simplified drawings of the net magnetization in motion.

Now you are ready to dig a little deeper into the matter and we continue with our MRI experiment and see what happens when we start to play around with the net magnetization.

To obtain an image from a patient it is not enough to put him/her 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.

Excitation

Before the system starts to acquire the data it will perform a quick measurement (also called pre-scan) to determine (amongst others) at which frequency the protons are spinning (the Larmor frequency). This centre frequency is important because this is the frequency the system uses for the next step.
Once the centre frequency is determined the system will start the acquisition.
This time we keep things real simple. No fancy pulse sequences. We come to that later.
For now we only send a radio frequency pulse into the patient and we look at what happens.

Let us assume we work with a 1.5 Tesla system. The centre or operating frequency of the system is 63.855 MHz. In order to manipulate the net magnetization we will therefore have to send an Radio Frequency (RF) pulse with a frequency that matches the centre 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.

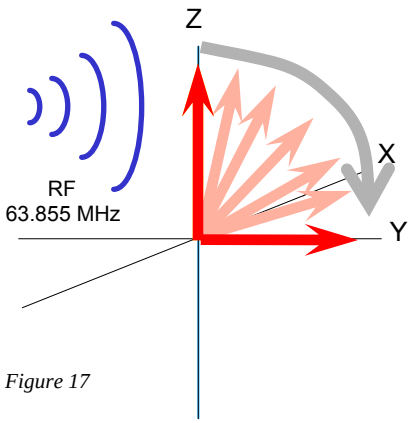


Figure 17

By sending an RF pulse at the centre frequency, with a 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 17*). (See how handy these vectors are. Without the vectors it would be quite impossible to draw this event).
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 amount of degrees we rotate the net magnetization. It is possible to

flip the net magnetization any degree in the range from 1º to 180º. For now we only use an FA of 90º.
This process is called excitation. That's it, applause!!

Relaxation

Now it becomes interesting. We rotated the net magnetization 90º into the X-Y plane. We could also say that we lifted the protons into a higher energy state, same thing. This happened because the protons absorbed energy from the RF pulse. This is a situation that the protons do not like. You could compare this with walking on your hands, it is possible but you don't like it for a long time. You prefer to walk on your feet. Same thing for the protons, they prefer to align with the main magnetic field or, in other words, they would rather be in a low energy state.
Now something happens that is referred to as Relaxation. The relaxation process can be divided into two parts: T1 and T2 relaxation.

T1 Relaxation

The protons want to go back to their original situation, called equilibrium. They do so by releasing the absorbed energy in the shape of (very little) warmth and RF waves.
In principle the reverse of excitation takes place. The net magnetization rotates back to align itself with the Z-axis.

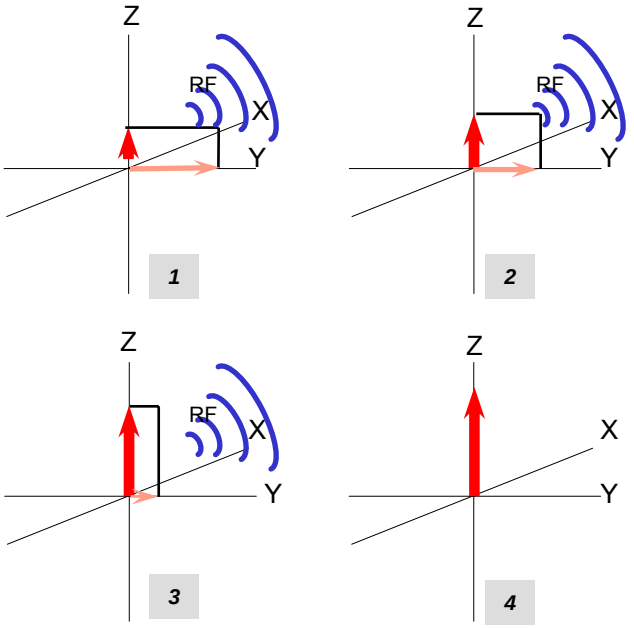


Figure 18

After the RF excitation pulse stops, the net magnetization will re-grow along the Z-axis, while emitting a radio-frequency waves (*Figure 18*).
T1 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.
T1 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 ones mind.

T1 Relaxation Curves

T1 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 ^{1}H 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 T1 relaxation can be depicted as shown in *Figure 19*.

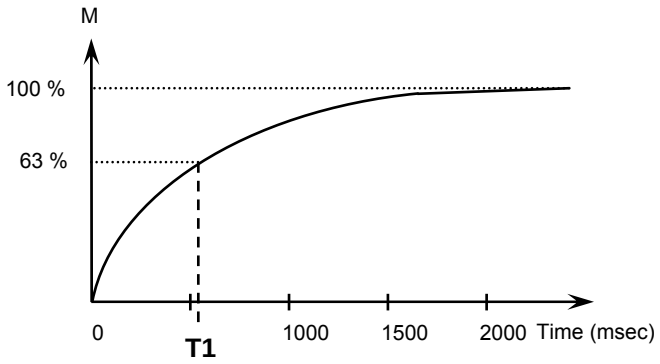


Figure 19

The curve shows at time = 0 that there is no magnetization in the Z-direction right after the RF-pulse. But immediately the M_z starts to recover along the Z-axis. T1 relaxation is a time constant. T1 is defined as the time it takes for the longitudinal magnetization (M_z) to reach 63 % of the original magnetization.
A similar curve can be drawn for each tissue. That's what **Damadian** and **Lauterbur** discovered many moons ago. Each tissue will release energy (relax) at a different rate and that's why MRI has such good contrast resolution.

T2 Relaxation

As said before, the relaxation process is divided into two parts. The second part, T2 relaxation, is slightly more complicated. I have found that people have difficulties in understanding this, but we'll have a jolly good go at it. Do not despair!!
First of all, it is very important to realize that T1 and T2 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. T1 relaxation describes what happens in the Z direction, while T2 relaxation describes what happens in the X-Y plane. That's why they have nothing to do with one another. I cannot emphasize this enough.

Phase and Phase coherence

Ever heard of Phase? Imagine this: if you see a group of soldiers marching along the road they all put their left leg forward at the same time. The sergeant tells them to do so: Left, Right; Left, Right, Left . . . Left . . . Left, Right. You could say that the group is walking in phase or in synchronization.

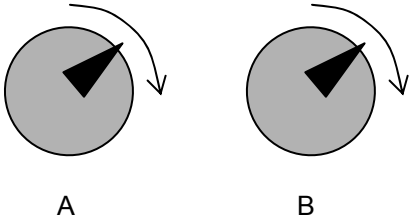


Figure 20

Another example: In *Figure 20* we see two wheels with an arrow. The wheels are rotating at the same speed. The arrows will therefore point in the same direction at any time. The wheels are said to be rotating In-Phase.

Let's go back one step and have a look at the net magnetization vector before we apply the 90º 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 90º RF pulse something interesting happens. Apart from flipping the magnetization into the X-Y plane, the protons will also start spinning *in-phase*!!

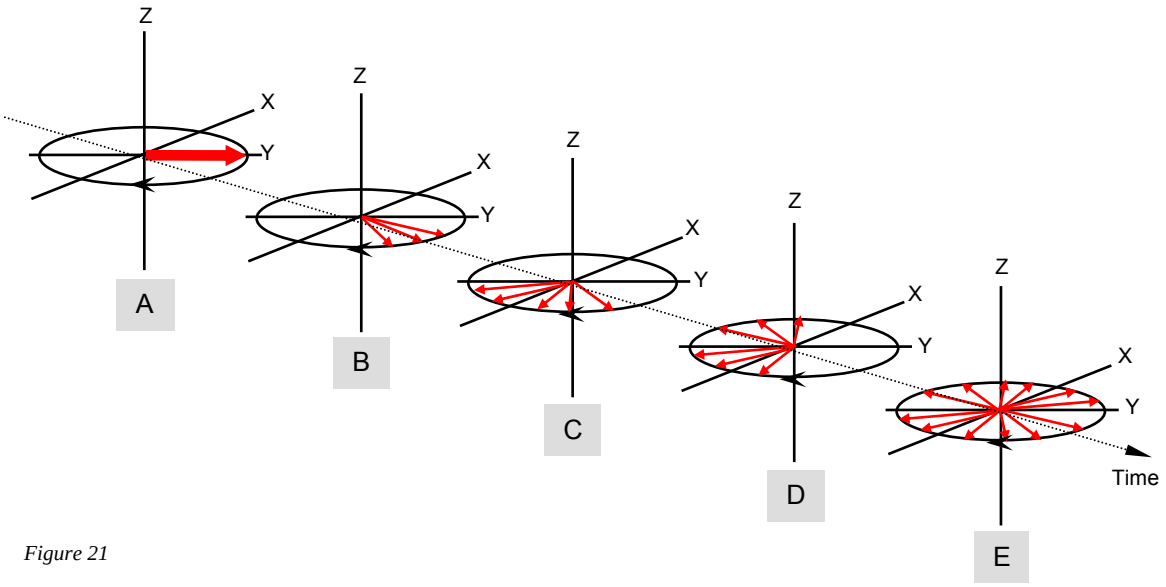


Figure 21

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 (*Figure 21A*). The vectors all point in the same direction because they are In-Phase. However, they don't stay like this.
Image this: I'm sure as a child you played the game where you stood real close behind each other and then you tried to walk. This only works when both of you put your left foot forward at the same time. Then you are walking In-Phase. At some time one of you would stumble and your feet would get tangled up resulting in a mini chaos with both of you walking in a different direction: you got Out-Of-Phase or you 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 21B, 21C, 212D*), but quickly that will increase until there is no more phase coherence left: there is not one vector pointing in the same direction anymore. (*Figure 21E*) In the meanwhile the whole lot is still rotating around the Z-axis in the X-Y plane.

This process of getting from a total *in-phase* situation to a total *out-of-phase* situation is called T2 relaxation.

T2 Relaxation Curves

Just like T1 relaxation, T2 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.

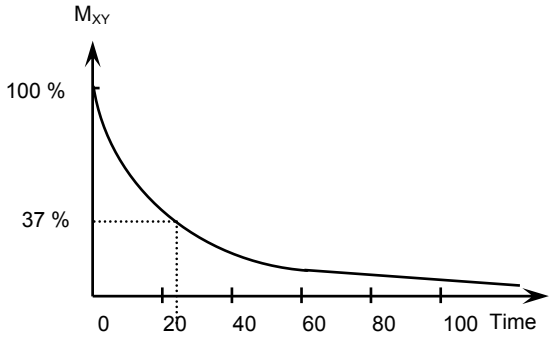


Figure 22

Also here we can draw a curve. (*Figure 22*)
Right after the 90º RF-pulse all the magnetization is "flipped" into the XY-plane. The net magnetization changes name and is now called M_{XY}. At time = 0 all spins are in-phase, but immediately start to de-phase. T2 relaxation is also a time constant. T2 is defined as the time it

takes for the spins to de-phase to 37% of the original value.
The rate of de-phasing is different for each tissue. Fat tissue will de-phase quickly, while water will de-phase much slower.
One more remark about T2: it happens much faster than T1 relaxation. T2 relaxation happens in tens of milli-seconds, while T1 can take up to seconds. (Have a look at the relaxation times table in the § Appendix).

T2 relaxation is also called spin–spin relaxation because it describes interactions between protons in their immediate surroundings (molecules).

Remember this:

- T1 and T2 relaxation are two independent processes, which happen simultaneously.
- T1 happens along the Z-axis; T2 happens in the X-Y plane.
- T2 is much quicker than T1

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 90º RF-pulse.

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.

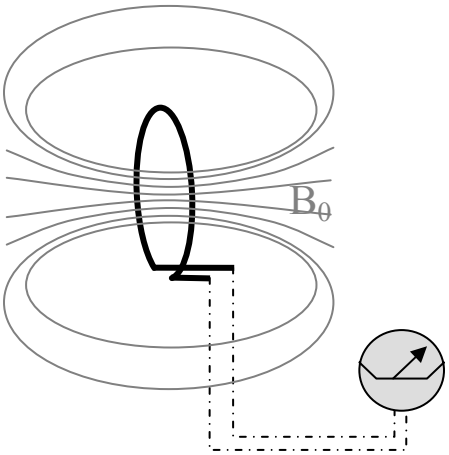


Figure 23

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*). B_0 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 B_0 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 24*.

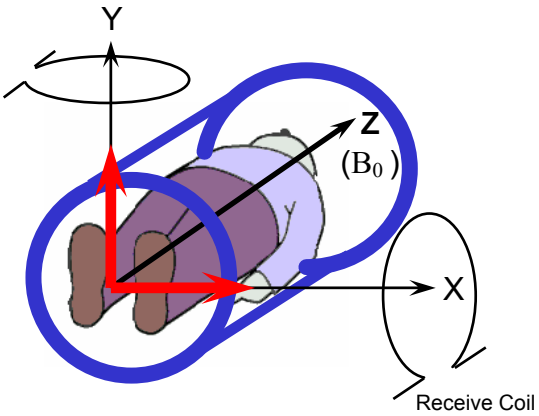


Figure 24

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 B_0, and then turn it a bit such that B_0 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_0.

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 a right angles from one another, have a 90º phase difference and both move in the same direction with the speed of light (*Figure 25*).

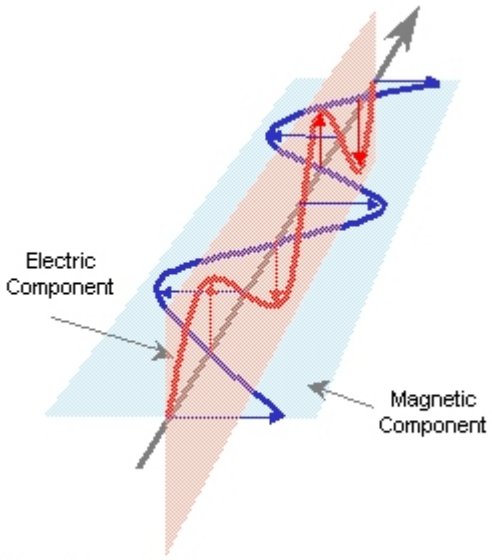
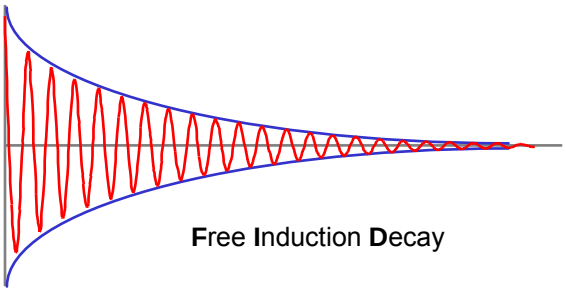


Figure 25

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

The story about positioning the coil at right angles to B_0 serves another purpose; it means that we can only receive signals from processes that happen at right angles to B_0, which happens to be T2 relaxation. T2 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.



Free **I**nduction **D**ecay

Figure 26

Consequently, the signal that is received is strong in the beginning and quickly becomes weaker due to T2 relaxation (*Figure 26*).

The signal is called: Free Induction Decay. The FID is the signal we would receive in absence of any magnetic field. In the presence of a magnetic field T2 decay goes much faster due to local (microscopic)

magnetic field inhomogeneity and chemical shift, which are known as T2* effects). The signal we receive is much shorter than T2. The actual signal decays very rapidly; in ± 40 milliseconds it's reduced to practically zero.
This poses a problem, as we will see later.

Computing and Display

The received signal is then fed into a computer and, amazingly, a quarter of a second later an image appears on the screen. There is an awful lot more to tell about the computing part, but that's beyond the scope of this story and, what's more, totally irrelevant (thank goodness, ☺). *Figure 27* shows the entire process graphically.

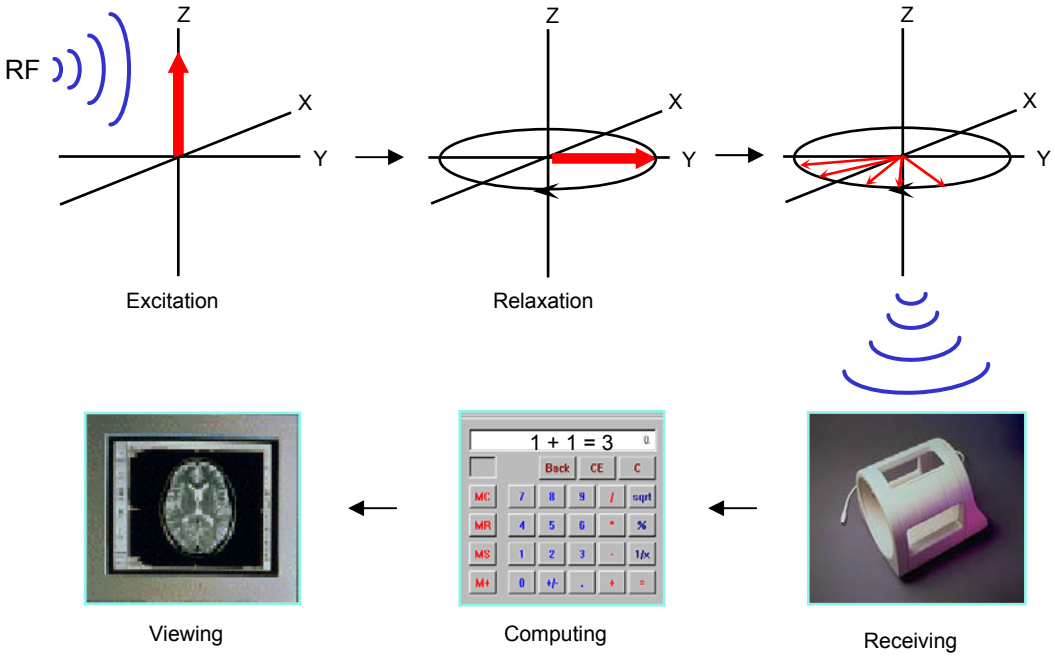


Figure 27

That's it folks. MRI in a nutshell. Obviously it is quite a bit more complicated, but this is basically what it boils down to.

More Physics

Now you know how MRI works. Good as it may be, you have only scratched the surface. This is a good time to get out your bucket and spade and dig a little deeper and find out what is hidden under that surface.

And when we start digging we run straight into a problem!

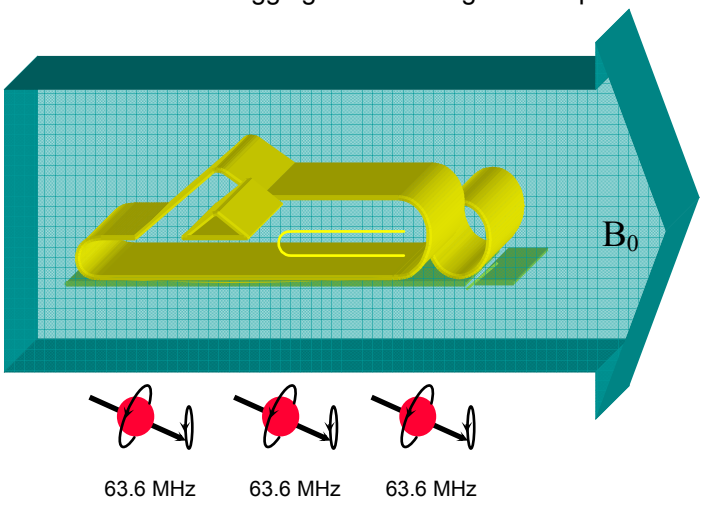


Figure 28

If we assume we have a 100% homogeneous magnetic field (which it isn't), then all the protons in the body would spin at the Larmor frequency *(Figure 28)*. This also means that all protons would return signal. How do we know whether the signal is coming from the head or from the foot? Well, we don't. If we would leave things as they are then we wouldn't get a pretty picture; certainly not one we would expect. It would have nothing but undecipherable blobs. The solution to our problem can be found in the properties of an RF-wave, which are: phase, frequency and amplitude. First we will divide the body up into volume elements, also known as: voxels. Then we are going to code the voxels such that the protons, within that voxel, will emit an RF wave with a known phase and frequency. The amplitude of the signal depends on the amount of protons in the voxel.
The answer to our problem is: Gradient Coils.

Gradient Coils

Gradient coils are a set of wires in the magnet, which enable us to create additional magnetic fields, which are, in a way, superimposed on the main magnetic field B_0. Sounds complicated, but it's not really.

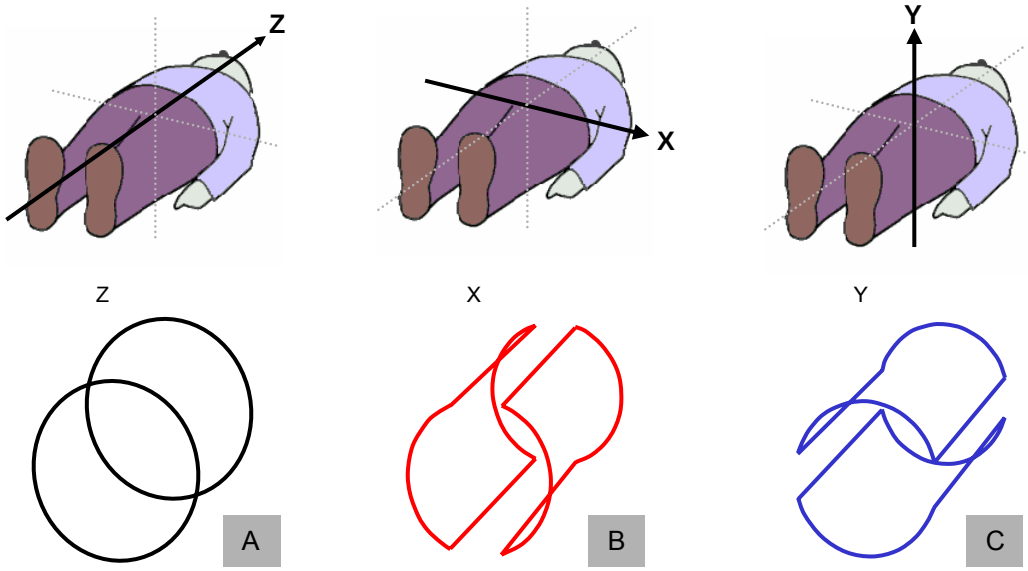


Figure 29

There are 3 sets of wires. Each set can create a magnetic field in a specific direction: Z, X or Y. When a current is fed into the Z gradient, then a magnetic field is generated in the Z direction (*Figure 29A*). The same goes for the other gradients. (*Figure 29B and 29C*)

Interesting detail: Everyone knows that MRI can make a lot of noise during acquisition. The magnetic field, which is generated, is very strong. Although the gradient coils are very tightly fixed in a kind of resin, the forces, exhibited by the gradient coil, are enough to make them vibrate, hence the noise.

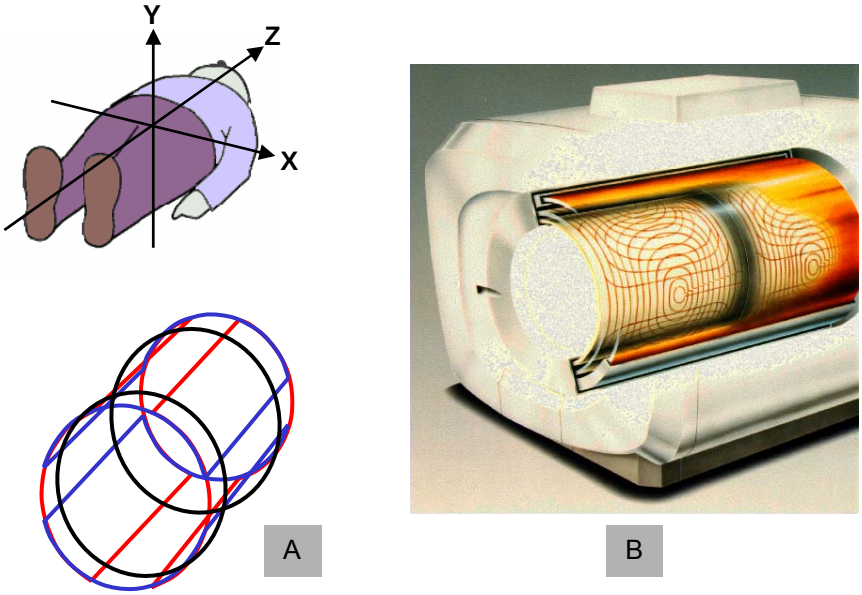


Figure 30

Figure 30A shows schematically how the 3 gradient coils form a cylinder. This cylinder is then placed in the magnet bore (*Figure 30B*).
Let's move on and discuss how the gradients are used to code the signal.

Signal Coding

First we make a few assumptions:

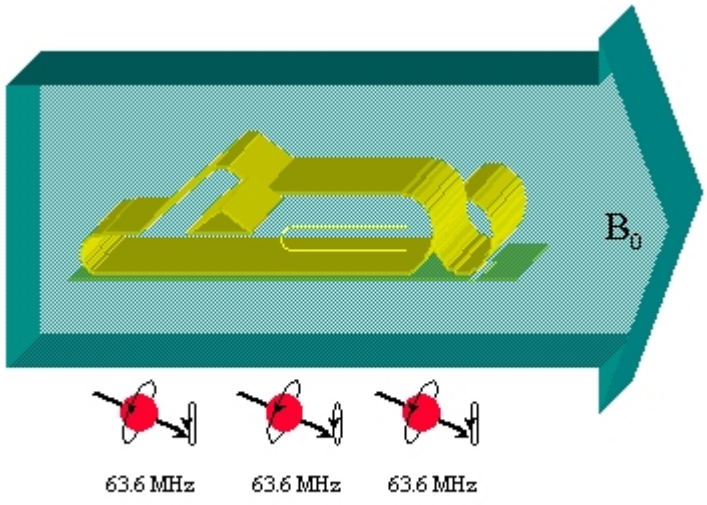


Figure 31

- We are going to make an axial image of the brain.
- We use a 1.5 Tesla magnet.
- 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 of the magnet, but it makes the story easier to explain).

When we put a patient in the magnet, all the protons, from head to toe, align with B_0. They spin at the Larmor frequency of 63.6 MHz. (*Figure 31*).

If we use a 90º 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.

Slice Encoding Gradient

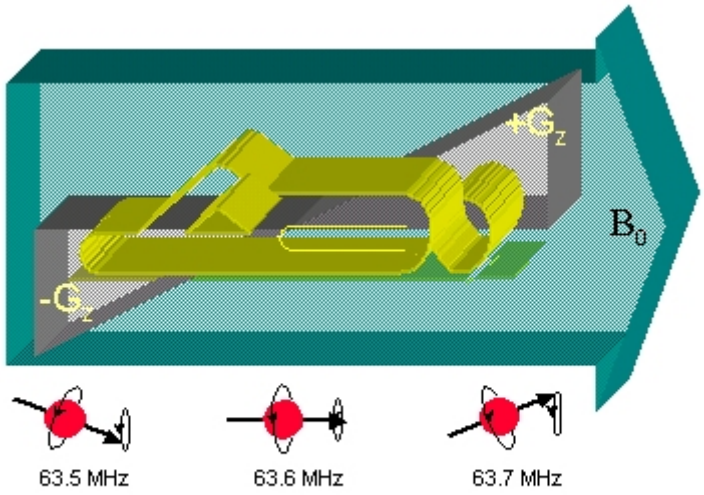


Figure 32

The Z-gradient is switched on. This will generate an additional magnetic field in the Z-direction, which is superimposed on B_0. The indication +Gz in *Figure 32* means there is a slightly stronger B_0 field in the head as there is in the iso-centre of the magnet. A stronger B_0 field means a higher Larmor frequency. Along the entire the slope of the gradient there is a different B_0 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-centre. The reverse goes for the protons in the feet. *Figure 32* shows that the protons in the feet now spin at 63.5 MHz, the ones in the iso-centre of the magnet still at 63.6 MHz and the ones in the head with 63.7 MHz.

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

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 (*Figure 33*).

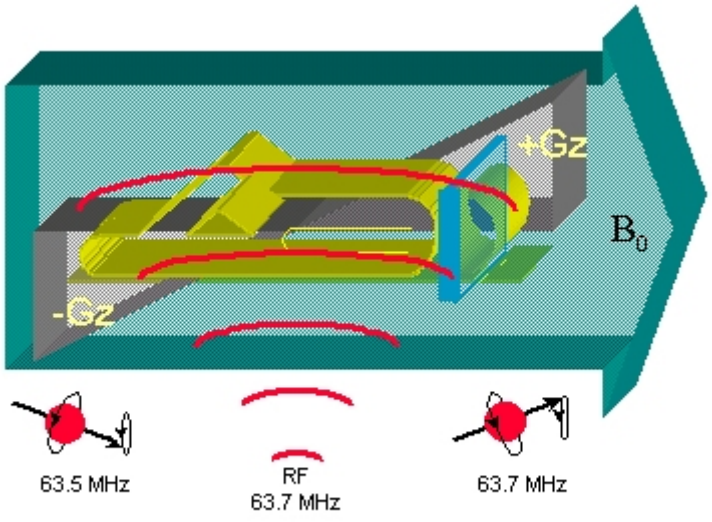


Figure 33

This is called Slice-Encoding or Slice-Selection. In this example Gz is the slice-encoding gradient.

If we would stop here and listen to the returned signal then we know that the signal comes from the single slice in the head.

We now know in one direction (Z-direction) where the signal is coming from. That is a big improvement.

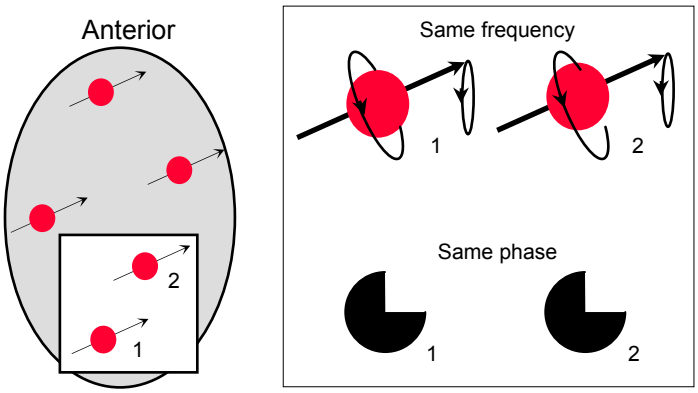


Figure 34

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 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.

Phase Encoding Gradient

In order to code the protons further the Gy 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.

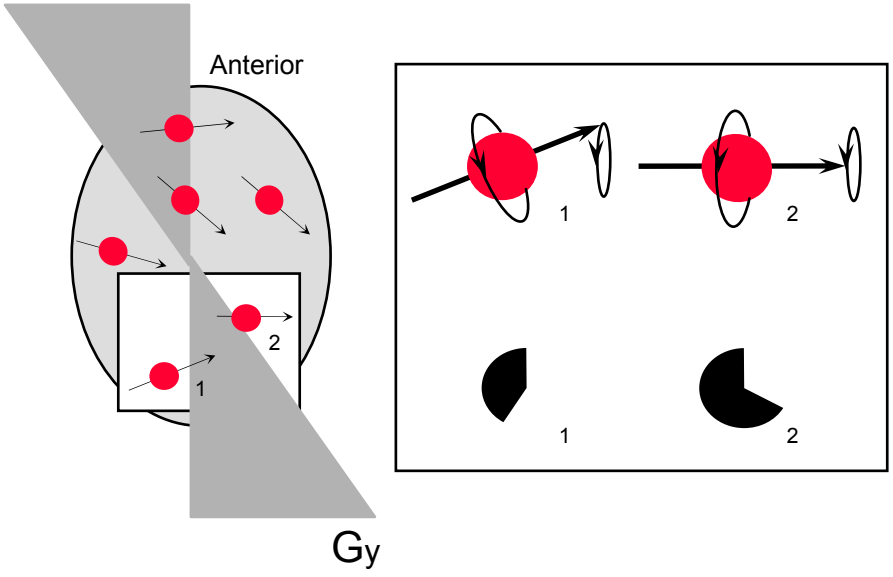
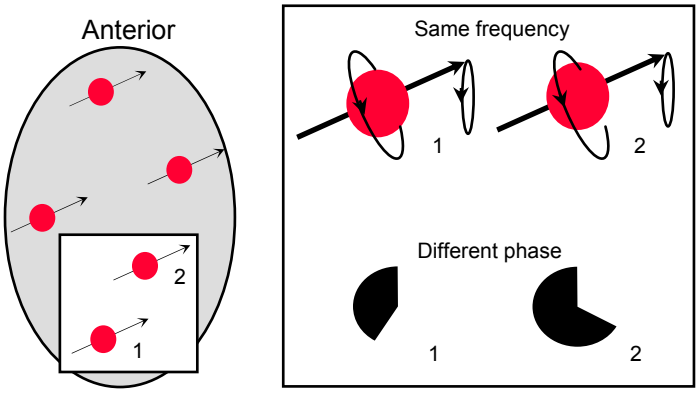


Figure 35

The effect is that the anterior protons will spin slightly faster than the posterior protons.

Because of this difference the protons do not spin In-Phase anymore. Looking at proton 1 and 2, we see that proton 1 has accumulated a slightly higher phase than proton 2 (*Figure 35*).



When the Gy gradient is switched off, each proton within the slice spins with the same frequency BUT each has a different phase (*Figure 36*)

This is called: Phase Encoding.

Figure 36

After this second encoding process we are a step further in our quest for pinpointing the exact origin of the signal.

We can determine two things:

1. The signal comes from a slice in the head. (Slice Encoding)
2. The signal contains a number of RF waves, which have the same frequency, but have different phases. It is possible to tell whether the signal comes from anterior or from posterior. (Phase Encoding)

All we need to do now is to do one more encoding to determine whether the signal comes from the left, the centre or the right side of the head.

Frequency Encoding Gradient

To encode in the left – right direction the third, and last, gradient (Gx) is switched on. This will create an additional gradient magnetic field in the left – right direction.

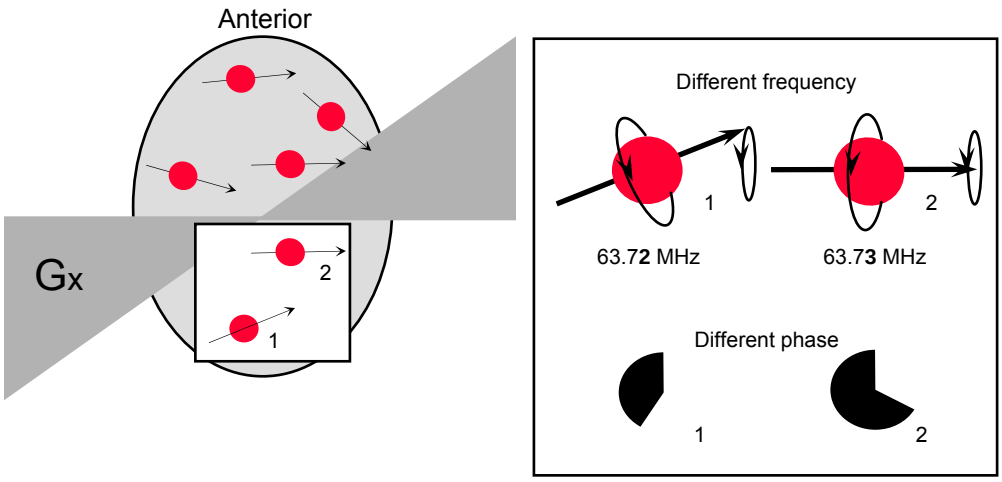


Figure 37

The protons on the left hand side spin with a lower frequency than the ones on the right. (*Figure 37*) 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, centre or right hand side of the slice.

Task completed!
We can pinpoint the exact origin of the signals, which are received by the coil.

Let's recapitulate all this and have a look what we've done during the entire process.

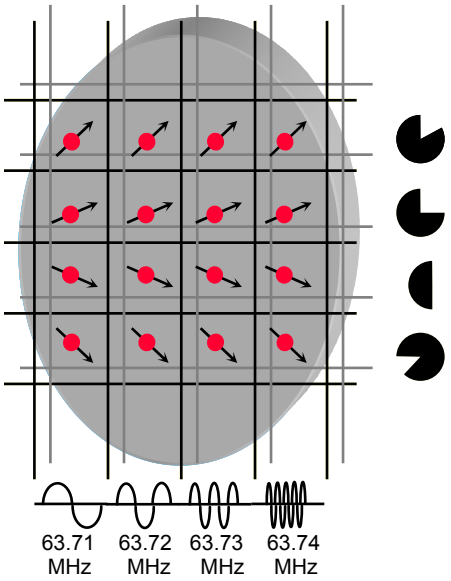


Figure 38

Figure 38 shows the end result:

1. The Gz gradient selected an axial slice.
2. The Gy gradient created rows with different phases.
3. The Gx gradient created columns with different frequencies.

As you can see small volumes (voxels) have been created. Each voxel has a unique combination of frequency and phase. The amount of protons in each voxel determines how strong (amplitude) the RF-wave is.
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 2 Dimensional Fourier Transform (2DFT), which enables the computer to calculate the exact location and intensity (brightness) of each voxel. *(It is way beyond the scope of this story to explain how Fourier Transformation works. However, you can compare FT with a prism, which brakes up 'white' light (MR signal) into the colours of the rainbow (image). We do FT all the time ourselves with our ears. We hear many different sounds (MR signal) at the same time. Our brain does a FT to pinpoint the origin and intensity of the individual sounds (MR image)*.

Remark: Phase Encoding can only be done one row at the time. In order to scan the whole slice the entire process of slice encoding, phase encoding and frequency encoding has to be repeated as many times as specified by the parameter Matrix $_{\text{phase encoding}}$ (Mx_{pe}). This also explains the necessity of the scan parameter Repetition Time (TR). More about scan parameters later.

Side Step: 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.

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 (see point 3).
3. Slew rate: as big as possible (min. TR, TE and ETS).

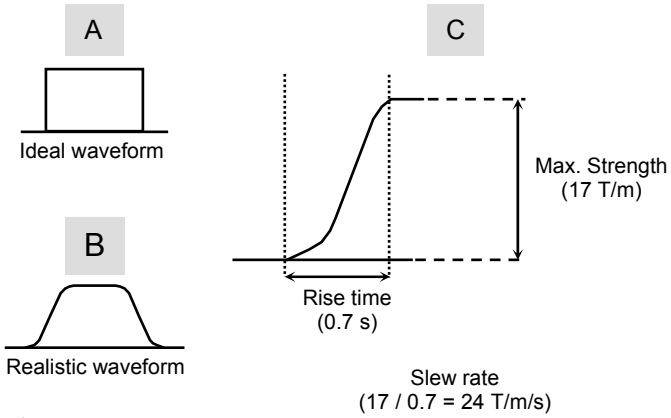


Figure 39

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_0, the computer system and the ease of use of the user interface.

Side Step: Slice Thickness

During the first step of signal coding the location of the slice in the body is determined by the Slice Encoding gradient (Gss). The slice thickness was for our example not important. In real life, however, the slice thickness *is* important.
The slice thickness is determined by two factors:

1. The steepness of the slope of the gradient field
2. The bandwidth of the 90º RF-pulse.

In *Figure 40A & 40B* the steepness of the gradient is kept the same, while the bandwidth of the RF-pulse is varied. Alternatively, *Figure 40C & 40D* show that varying the steepness of the

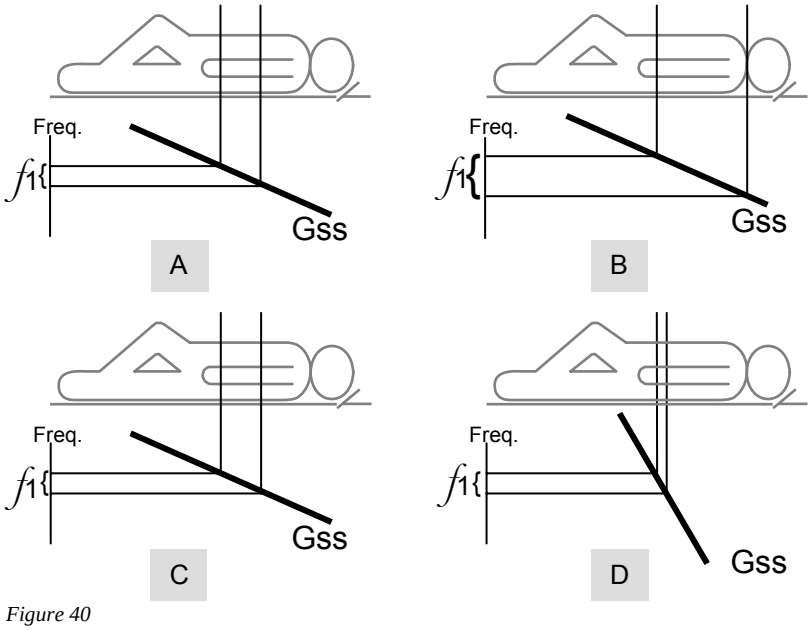


Figure 40

gradient, while keeping the RF-pulse bandwidth the same, can also change the slice thickness. In practice, the slice thickness is determined by a combination of both gradient steepness and RF-pulse bandwidth.

Even More Physics

Don't put away your bucket and spade just yet. We're going to dig a little bit deeper.
As said before, the physics of MRI is a highly complex matter. The physics can be divided into various parts, which have apparently got nothing to do with one another and, yet, they are all linked together.
So far, we have talked about the general picture and how to code the signal. In the next section we go to a different part of the story, where we learn about how the signal is acquired and stored before becoming an image.

A journey into *k*-Space

People write books about *k*-space alone, it's that complex. When I first heard about *k*-space and the story that came with it, all I could mutter was 'uh, say again'. What is so difficult about *k*-space? Honestly, I don't know. Probably the facts that it is not tangible, kind of hard to image, pretty much like the sixth sense. The truth is I was never told the right story until I read Moriel NessAiver's book (see references). He has this beautiful, almost romantic description about what *k*-space is:

"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*. Same things. The reason why the phrase *k*-space is used and not the others are because everybody uses it in the literature.

Question: Why is *k*-space so important?
Answer: It helps us to understand how an MRI image is acquired and how various pulse-sequences work.

Here goes. Hold on to your hat.

Figure 41 shows a square. This is a representation of *k*-space, matrix, time-domain or whatever you would like to call it. We see two lines, X and Y, which divide the box such that both left, right and top, bottom are symmetrical. In this box we are going to put our MRI data before it gets reconstructed into an image.

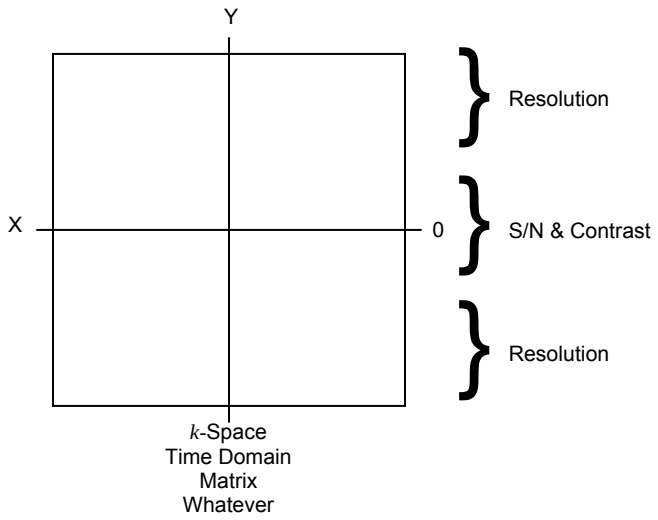


Figure 41

The acquired data (also known as raw data) is put into this box such that signals with low frequencies go in the centre and those with high frequencies are spaced around the centre.

Signal with low frequencies contain information about signal and contrast, while high frequencies contain information about spatial resolution (sharpness).

This is still a bit difficult to imagine, but that may change if you look at *Figure 42*.

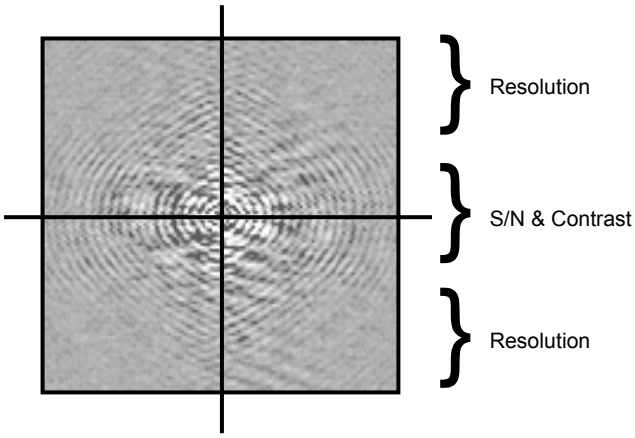


Figure 42

Raw data can be reconstructed in two ways. *Figure 42* shows one way of reconstruction we are not very familiar with. This is what we call a raw data image. It is just a different representation of the image data. The image shows very clearly that the data is spaced around the centre. You can also see that the centre contains high and low signal as well as contrast information. Spaced around the centre you see 'rings', which make up information regarding spatial resolution. Furthermore, it can be seen that *k*-space is symmetrical both from left to right and from top to bottom.

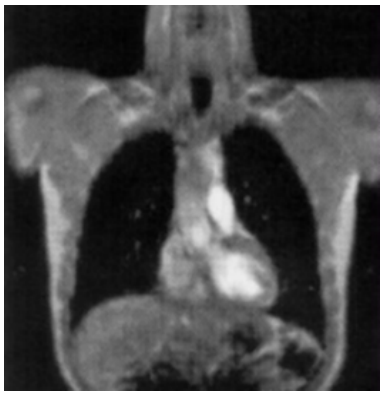
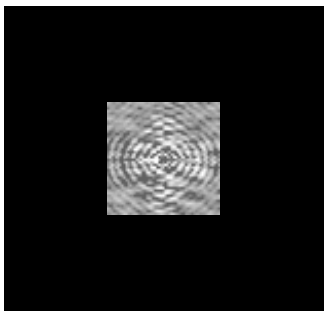


Figure 43

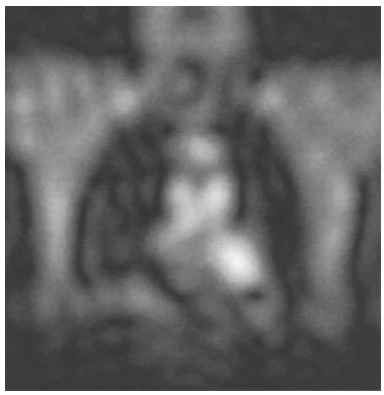
The other way of reconstructing raw data will give us an image, which we recognize immediately (*Figure 43*). This image is reconstructed from the same raw data set.

Engineers use the raw data image (*Figure 42*) to obtain more information regarding image artifacts. Image artifacts are usually caused by 'wrong' frequencies.

To illustrate the fact that information about signal/noise and contrast is stored in the centre of *k*-space we can do the following experiment. Have a look at *Figure 44*. Here we only reconstructed the central part of *k*-space (*Figure 44A*). The resulting image (*Figure 44B*) shows contrast, but the image is very blurred. This is because we left out the information regarding spatial resolution, which is stored in the outside of *k*-space.



Central part



A B

Figure 44

We can do the same thing, but this time we only reconstruct the outside of *k*-space (*Figure 45A*). The resulting image (*Figure 45B*) shows sharp contours, but almost no contrast information.

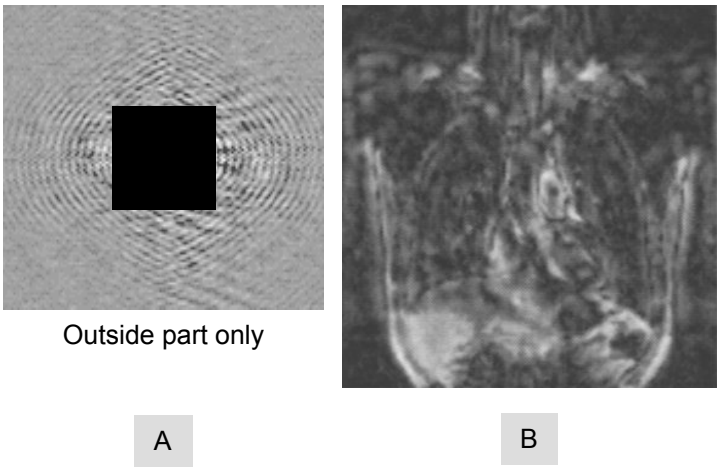


Figure 45

With this out of the way, let's take it a step further, and discuss how *k*-space is filled during an acquisition. Here we are going to piece some things together which we have discussed so far. This is going to be a bit complex, so hang in there.

Filling *k*-Space

In the previous paragraph I mentioned briefly that Phase Encoding could only be done one line at a time. We have to repeat the whole process of excitation, phase encoding and such as many times as we specified with the parameter MX_{PE}. *K*-space is, therefore, also filled one line at a time. *Figure 46* tells the whole story.

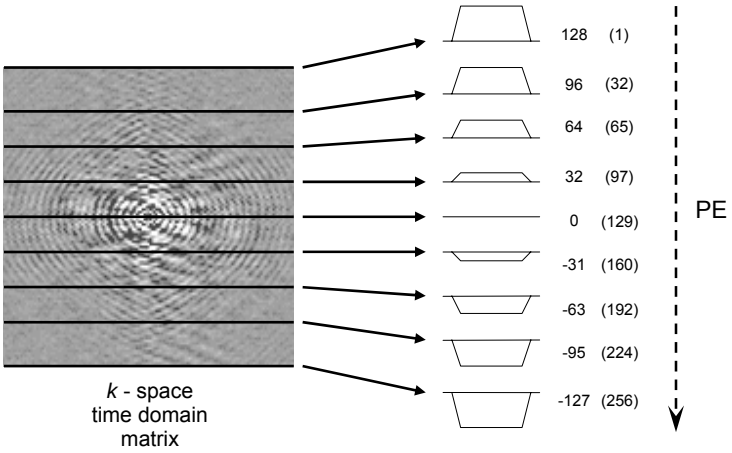


Figure 46

Let us assume we are scanning with a Matrix of 256 x 512. The first number refers to the matrix in the "phase encoding direction" (MX_{PE}), while the last number refers to the matrix in the "frequency encoding or read-out direction" (MX_{RO}).
The funny graphics, which the arrows point to, indicates the gradient power and polarity (+ or -). The numbers between brackets indicate the line number in *k*-space: the first line is 1; the last line is 256. Each line will be made up of 512 points as specified by MX_{RO}.

In the first repetition of our sequence, a + gradient with a power of 128 will be applied and the 1st line of *k*-space will be filled. In the second repetition, a + gradient with a power of 127 will be applied and the 2nd line of *k*-space will be filled. In the 129th repetition, no gradient is applied and the 129th line of *k*-space will be filled. In the 160th repetition, a – gradient with power –31 will be applied and the 160th line in *k*-space will be filled, and so on until the entire *k*-space is filled. Because we choose the phase encoding matrix (MX_{PE}) to be 256, the scan will be repeated 256 times. Alternatively, if we choose 192 for the MX_{PE}, then the scan will be repeated 192 times and *k*-space will, therefore, have 192 lines in the phase direction.

I hope all this makes some sense.

k-Space Symmetry

In our scan example we filled the entire *k*-space from top to bottom, starting with the 1st line and finishing with the 256th line. This is indeed the way things are done in a routine scan.
I mentioned before that *k*-space is symmetrical in both directions. We can use this symmetry to our advantage.
We do *not need* to fill *k*-space from top to bottom! When we fill *k*-space slightly more than 50%, it is then possible to fill the missing lines with ones we already acquired.

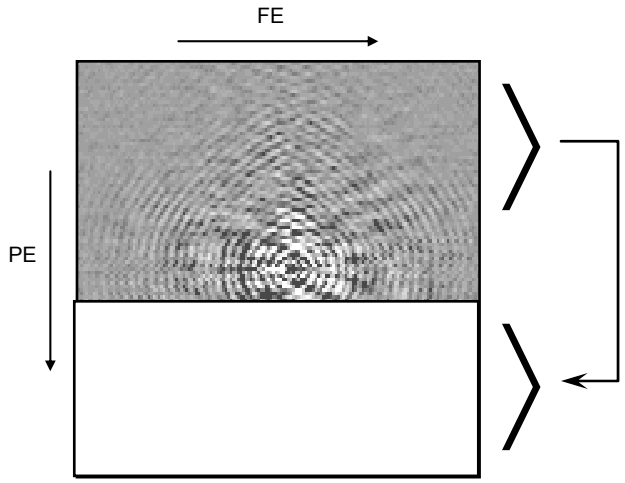


Figure 47

Figure 47 shows how this works. About 57% of *k*-space has been filled. The bottom half is filled with data from the top half. The big advantage is the reduction in scan time, because we only need to repeat the scan, let's say, 146 times. This is an enormous time saver, when you realize you could bring the scan time down from 6 minutes to slightly more than 3 minutes.

This technique is known as: Half Fourier Imaging, Half Scan, Partial Scanning or Advanced Fourier Imaging, depending on the manufacturer.

However, nothing comes for nothing in life and there is a down side to this trick: the resulting image is somewhat blurred. The reason is that k-space is not perfectly symmetrical, because we don't live in a perfect world. This trick is only applied when we need very fast scanning in applications such as some cases of MR Angiography or Perfusion/Diffusion scanning.

k-Space Filling Techniques

So far we filled *k*-space from top to bottom, but there are more ways to do it.
Figure 48 shows a few examples.

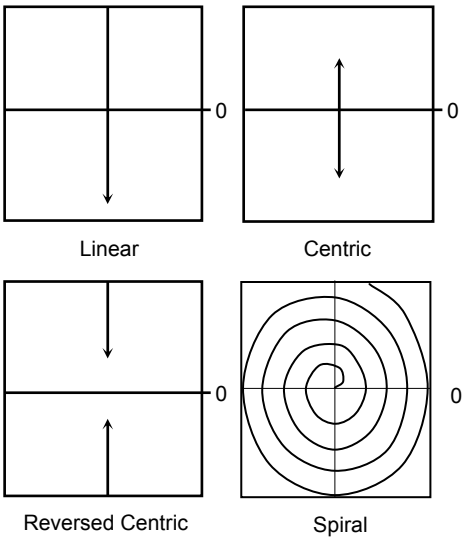


Figure 48

Linear
The method we used so far is also knows as linear *k*-space filling.

Centric
As the name implies, centric *k*-space filling starts in the centre. This means that the data from the 1st repetition of our scan is not put in the 1st line of *k*-space, but at the zero line. This is useful when you want to store contrast information first as would be the case when you do a contrast enhanced MR angio.

Reversed Centric
This method is not used very often as far as I know. It is also beyond me what the advantage would be.

Spiral
This is a special case. This method is used with very fast scan techniques such as in Single Shot Echo Planar Imaging. In one acquisition the whole of *k*-space is filled. The disadvantage is the poor spatial resolution. Usually a matrix of 64 x 64 can be filled. In order to get higher resolution one could use Multi Shot EPI, which would allow for a 256 x 256 matrix. Another disadvantage is its high sensitivity for magnetic field inhomogeneity.

Practical Physics I

We have a come a long way already. We have discussed various aspects of MRI physics. In the next section I will discuss image contrast and a number of pulse sequences, which are commonly used in MRI. Without pulse sequences we can't do MRI. Our life depends on it in terms of which kind of image contrast we want to see or, even, which kind of pathology we want to detect. Understanding what a pulse sequence is and how it influences the image is vitally important.

Pulse Sequences

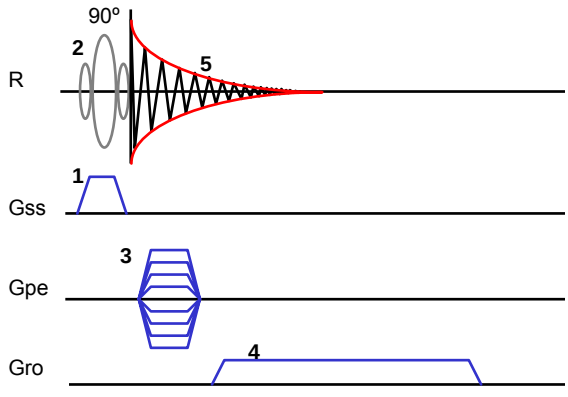


Figure 49

A pulse sequence is a sequence of events, which we need to acquire MRI images. These events are: RF pulses, gradient switches and signal collecting.
Figure 49 shows a "sequence diagram" in which the order of the events are shown schematically. These diagrams can be found in any book about MRI physics, so you better get used to them ☺.
Let's go back to our first experiment. We started of with (**1**) switching on the Slice Select gradient (G_{SS}). Simultaneously (**2**) a 90º RF-pulse was given to 'flip' the net-magnetization into the X-Y plane. Then (**3**)

the Phase Encoding gradient (G_{PE}) was switched on to do the first phase encoding. Then (**4**) the Frequency Encoding or Read Out gradient (G_{RO}) was switched on during which (**5**) the signal, the Free Induction Decay (FID), was sampled.

This is a very simple and basic sequence. We also saw that the signal dies out very quickly. In the early days that was a problem. The hardware could not be switched quick enough to sample the entire signal. They could only sample the last part of the signal when most of the signal was gone. The resulting image showed it! It was signal starved. In order to improve the amount of signal the engineers came up with a brilliant solution.

Spin Echo Sequence

After the 90º-excitation pulse the net-magnetization is in the X-Y plane. It immediately starts to dephase due to T2 relaxation (spin-spin interactions). It is because of this dephasing that the signal drops like a stone. Ideally, we would like to keep the phase coherence because this gives the best signal. The brilliant solution the engineers came up with is this: a short time after the 90º RF-pulse a second RF-pulse is given. This time it is an 180º pulse. The 180º pulse causes the spins to *re*phase. When all the spins are rephased the signal is high again, and when we make sure we sample the signal at this instant we'll have a much better image. *Figure 50* shows it better.

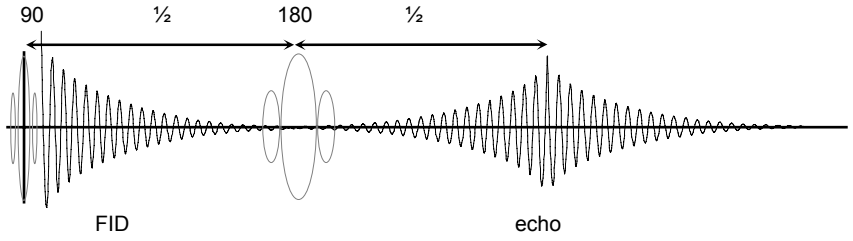


Figure 50

The signal we sample is called: an Echo, because it is "rebuilt" from the FID.
Notice that the 180º *rephasing* pulse is exactly in the middle of the 90º pulse and the echo.

There is a book (highly recommended, see reference) called: "MRI made easy… Well almost" by Schering, Germany, in which you can find a really nice analogy of rephasing:

Imagine a number of runners on a racetrack. When the whistle blows they all start to run (dephasing). Obviously they all run at the different speeds and after, let's say 30 seconds, the fastest one will be way ahead of the slower one. Then the whistle blows for the second time. The runners were instructed to turn around, *without losing speed*, at the second blow of the whistle (180º RF pulse). The fastest runner, now way behind the slower one, will catch up with the slower one (rephasing). After another 30 seconds they all arrive at the starting point *at the same time* (echo).

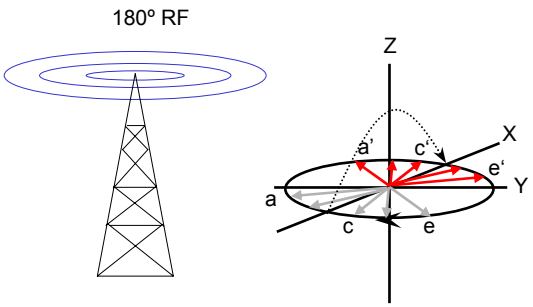


Figure 51

The effect of the 180º RF pulse is called: rephasing.
Figure 51 shows how this works. The spin system is mirrored around the Y-axis. Note that the rotation direction in the X-Y plane does not change.

So, what do we have? Let's have a look at *Figure 52*.

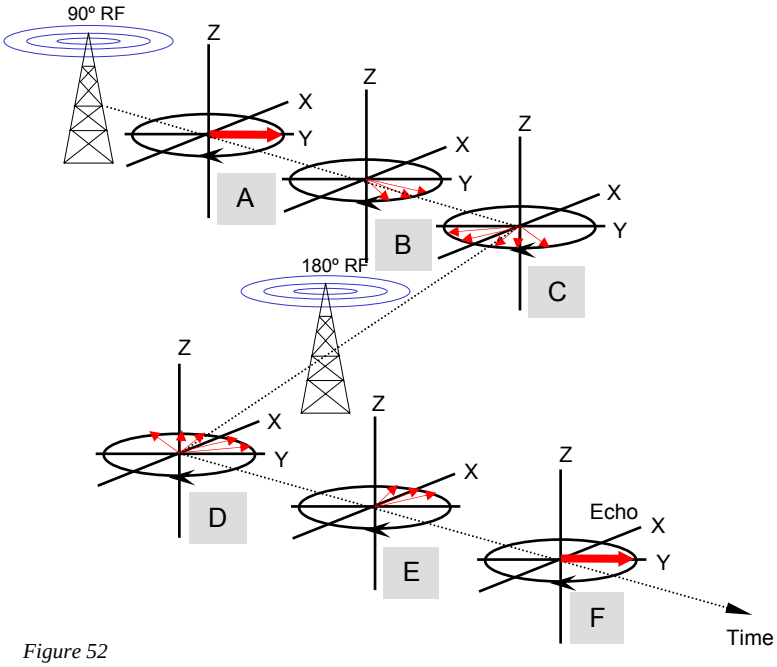


Figure 52

A. It starts with a 90º excitation pulse. The magnetization is flipped into the X-Y-plane.
B. Immediately the spins dephase…
C. The spins dephase a bit more… then a 180 rephasing pulse is given.
D. The spins are mirrored around the Y axis.
E. The spins rephase until…
F. The spins are in phase again creating an "echo".

This is what is known as a Spin-Echo sequence.
As with everything in MRI, the spin-echo sequence is a compromise:

Advantages:

- the signal is strong
- compensation for local field inhomogeneities: less artifacts.

Disadvantages:

- It takes time to do the rephasing step. This will increase the total scan time.
- It increases the amount of RF one has to put into the body (not that it's dangerous, but there are certain limits).

In spite of the increased scan time and the amount of RF the spin-echo sequence is widely used and has become the routine sequence in MRI.

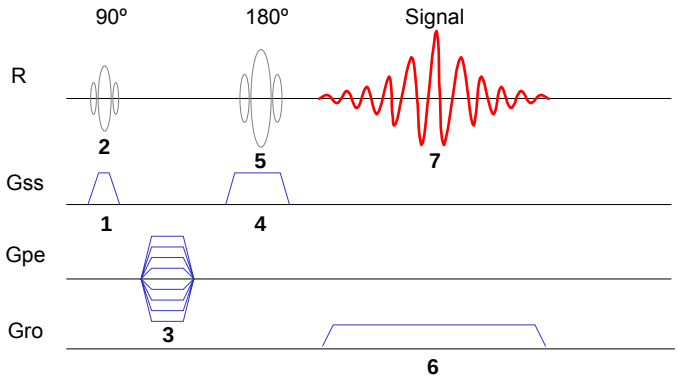


Figure 53

Figure 53 shows the pulse sequence diagram. Notice that during the 180º rephasing pulse the slice select gradient (G_{SS}) is switched on.
We start of with (**1**) switching on the Slice Select gradient (G_{SS}). Simultaneously a 90º RF-pulse (**2**) is given to 'flip' the net-magnetization into the X-Y plane. Then the Phase Encoding gradient (G_{PE}) (**3**) is switched on to do the first phase encoding. G_{SS} (**4**) is switched on again during the 180º rephasing pulse (**5**), so the same protons which were excited with the 90º pulse (**2**) are affected. Then the Frequency Encoding or Read Out gradient (G_{RO}) (**6**) is switched on during which (**7**) the signal is sampled.

At this moment in time I can introduce a few sequence parameters.

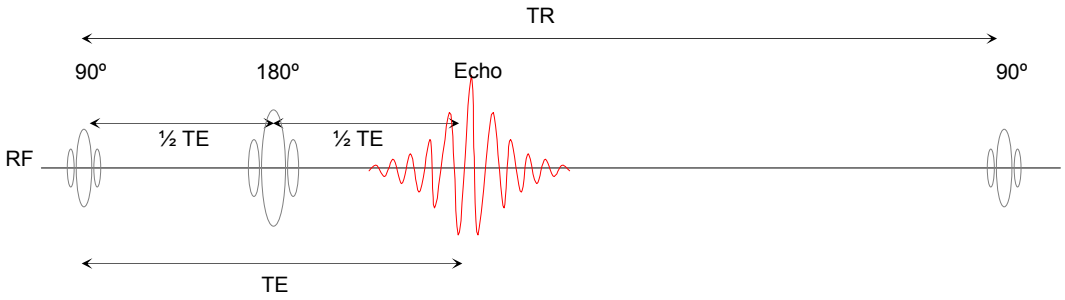


Figure 54

TR (Repetition Time). As stated before, the whole process must be repeated as many times as the matrix in the phase encoding direction is deep. TR is the time between two 90º excitation pulses. In regular SE sequences the TR can be anything in the range of 100 to 3000 milliseconds.
TE (Echo Time). This is the time between the 90º excitation pulse and the echo. The TE can be anything in the range of 5 to 250 milliseconds.
FA (Flip Angle). Refers to the amount of degrees the net-magnetization is flipped into the X-Y plane. It has nothing to do with the 180º rephasing pulse. The FA in a normal SE sequence is always 90º, however, in modern SE sequences this can be varied as well. FA's of 70º and 120º are quite common, although the FA can be choosen between 1º and 180º.

Multi Slicing

There is another reason why we need a TR. Remember that there are two relaxation processes going on at the same time: T1 and T2 relaxation. We have also seen that T1 relaxation takes much longer than T2 relaxation. When we want to repeat the scan for the next phase encoding we have to make sure that there is enough magnetization along the Z-axis. In other words, we have to allow for T1 to take place. If we don't allow enough time for T1 to happen we shall not have enough magnetization available for the next repetition and we would lose signal. Furthermore, as we will see later, the TR is an important parameter in influencing the contrast in the image.

An example: Let us assume we are making a scan of the brain.

We need 18 slices to cover the whole brain from the apex to the base of the skull.
We scan with a TE of 30 milliseconds.
We choose a TR of 540 milliseconds to have enough magnetization available for the next repetition.
We use a matrix of 256 x 512 (MX_{PE} = 256)

The time we need to scan the whole brain can be calculated as follows:

(TR x MX_{PE} x Number of slices) ÷ 60000

(540 x 256 x 18) ÷ 60000 = 41.4 minutes

That's a long scan time! Fortunately, it does not need to be so long. Let's have a closer look at the TR. To acquire the data we only need 30 milliseconds. We use a TR of 540 milliseconds. This means that we spent an awful long time (510 ms.) doing nothing while T1 relaxation is taking place. The waiting time is known as "dead time".
It is possible to use this dead time to our advantage. As soon as the first repetition is finished, the next repetition is started, but this time G_{SS} is shifted such that a slice next to the first slice is selected. When the first repetition of this second slice is acquired G_{SS} is shifted again to select a third slice and so on (*Figure 55*). After 540 milliseconds it is time to do the second repetition of the first slice. Right after this the second repetition of the second slice can be done followed by the second repetition of the third slice and so on.

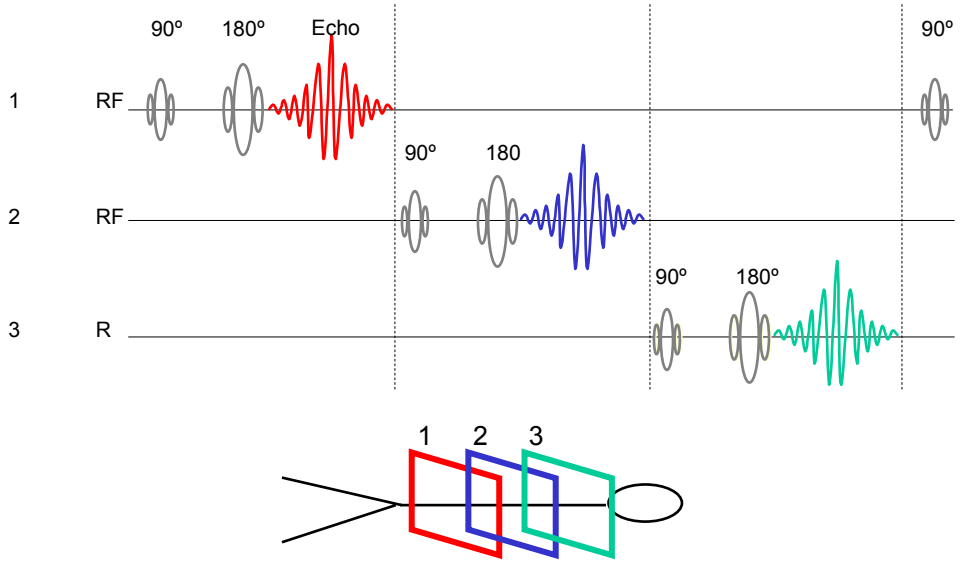


Figure 55

So, in 540 ms we are able to scan 18 lines of 18 different *k*-spaces generating 18 different images.
If we calculate the scan time again it works out that we only need 540 x 256 = 2.3 minutes. Now, that's a scan time we can live with. Especially when after this time the whole brain is imaged.

This trick is called Multi-Slicing. It is used in just about every scan we make so as not to waste any time. Just imagine when multi slicing would not be possible. MRI would be even slower than the Internet ☺.

Multi Echo Sequence

So far we used one echo in our sequence. By repeating the sequence we filled one *k*-space and this generated one image. It is, however, possible to acquire more echoes in one sequence.

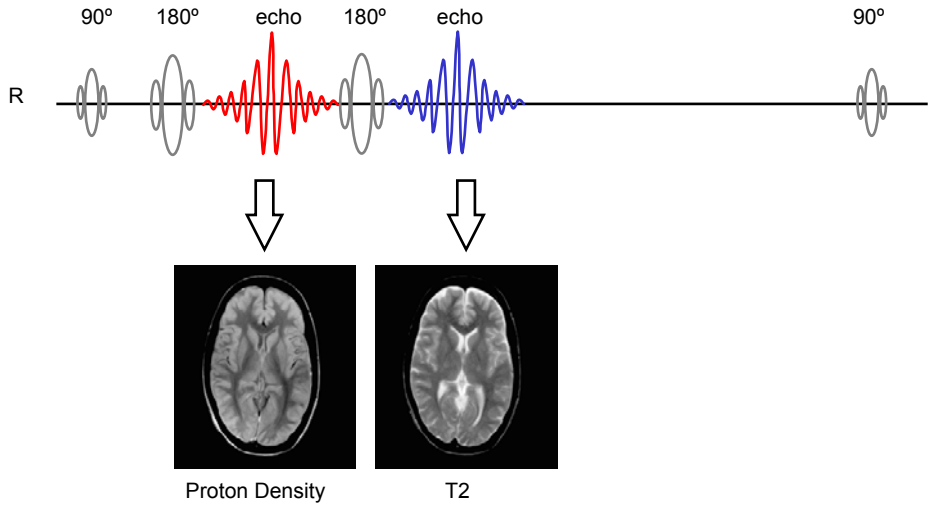


Figure 56

When an 180° rephasing pulse is applied we saw that the spins are rephased until they were all in phase again, but it doesn't stop there. What happens next is that the spins will, once again, start to dephase due to T2 properties. So, once more, we can apply a (second) 180° rephasing pulse to rephase the spins until they create a second echo. When we sample the second echo we put it in a second *k*-space. When all the lines of both *k*-spaces have been filled we end up with two different images. The second image has a different contrast because the TE is different.
The first image is a so-called Proton Density (see page 38 for explanation) image, while the second image is called a T2 image.
If we look at the images in *Figure 56*, we see this contrast difference. The cerebro-spinal-fluid (CSF) in the PD image is dark, while it is bright in the T2 image.
This type of sequence is called a Double-Echo Spin Echo sequence or, more commonly used, a SE T2 sequence.

This technique can be combined with Multi-Slicing bearing in mind that for a T2 sequence a very long TR is used (2000 ms or more). The long TR is necessary to allow for complete T1 relaxation of water.

Image Contrast

Before we move on to other pulse sequence techniques it is a good time to discuss image contrast.
We have seen that there are two relaxation processes, T1 and T2, going on at the same time. The image contrast is highly dependent on these relaxation processes. Image contrast depends on how much of each process we allow to happen. An example might be useful here:

T1 Contrast

Assume we scan with the following parameters: TR 600 and TE 10.

We allow for T1 relaxation to take place for 600 milliseconds and, more important, T2 relaxation only for 5 milliseconds (10÷2).

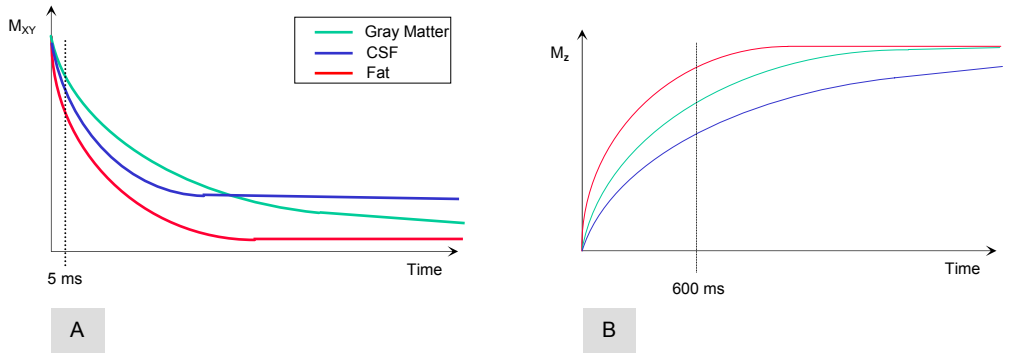


Figure 57

When we look at *Figure 57A*, we see that after 5 ms. hardly any dephasing has taken place. We receive a lot of signal from all tissues. The image contrast is, therefore, very little influenced by T2 relaxation.
In *Figure 57B* we see that after 600 ms. not all tissues have undergone complete T1 relaxation. Fat is nearly there, but CSF has still a long time to go. So, for the next excitation the net magnetization vector of the CSF spins, which can be flipped into the X-Y plane is small. This means that the contribution from CSF to the overall signal will be small too. In short, the image contrast becomes dependent on the T1 relaxation process. In the final image CSF will be dark, fat will be bright and gray matter will have an intensity somewhere in between.

In this case we say that the image is “T1 weighted” because the contrast is more dependent on the T1 relaxation process.

T2 Contrast

In another example we use the following parameters: TR 3000 and TE 120. Now we allow T2 relaxation to happen for 60 ms. (120÷2). As we can see from *Figure 58A* most of the tissues have dephased and won’t produce that much signal. Only CSF (water) has still some phase coherence left. Here the TE is the dominant factor for the image contrast.

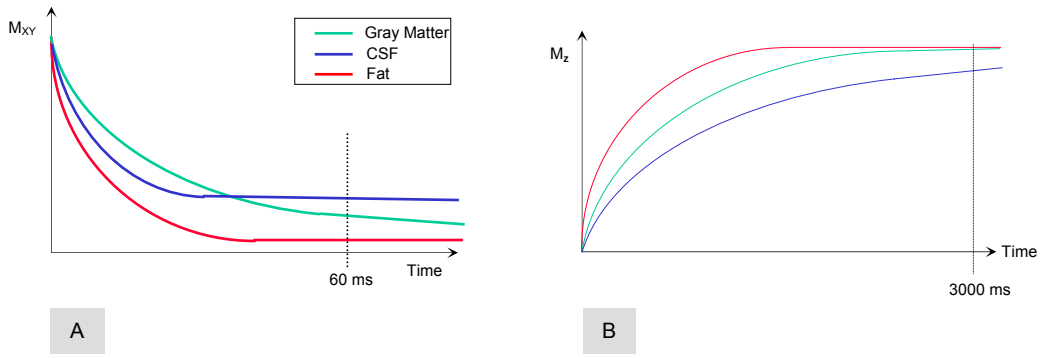


Figure 58

Figure 58B shows that practically all tissues have undergone complete T1 relaxation.
The long TR of 3000 ms does not contribute much to the image contrast. The 3000 ms. are only needed to allow CSF to recover completely before the next excitation.
In our image we'll see CSF bright, while the other tissues show up in various shades of gray.

In this case we say the image is "T2 weighted" because we allowed for T2 to happen for a "long" time.

Proton Density Contrast

There is one more type of image contrast called Proton Density.
Now we choose the parameters: TR 2000 and TE 10. Again we allow T2 relaxation to happen for only 5 ms., which means that T2 relaxation contributes very little to the image contrast.
With a TR of 2000 ms. the net magnetization of most tissues will have recovered along the Z-axis. The image contrast in PD images is neither dependent on T2 relaxation, nor T1 relaxation.
The signal we receive is completely dependent on the amount of protons in the tissue: few protons means low signal and dark in the image, while many protons produce a lot of signal and will be bright in the image.

It is important to understand that all images have a mix of T1 and T2 contrast. It just depends on how much T2 relaxation one does allow to happen. In SE sequences the TR and TE are the most important factors for image contrast.

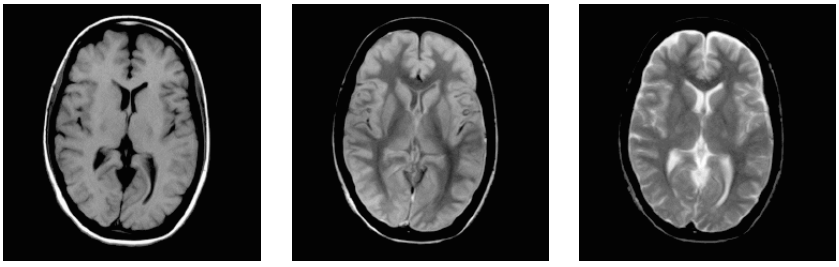


Figure 59

Figure 59 shows examples of various image contrasts; T1 weighted, Proton Density and T2 weighted. Notice the differences in signal intensity of the tissues. CSF is dark in T1, gray in PD and bright in T2.
It is quite amazing that only two parameters, TR and TE, can create so many different contrasts, even more so when we choose different values for these two parameters! I don't want to discourage you too much, but there are two more parameters, which influence image contrast.
Maybe now you understand why MRI is slightly more complex than CT scans.

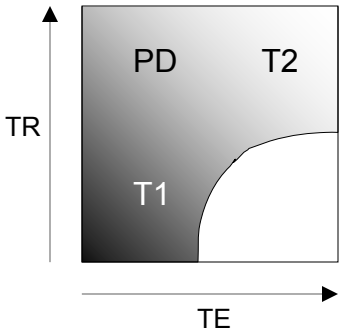


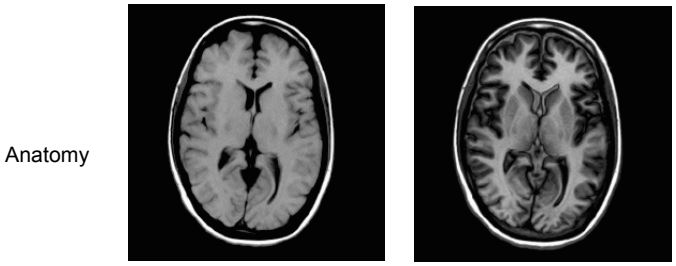
Figure 59A shows in a diagram how TR and TE in a SE sequence are related in terms of image contrast. A short TR and short TE gives T1 weighted contrast. A long TR and a short TE gives PD contrast. A long TR and long TE gives T2 weighted contrast.

Figure 59A

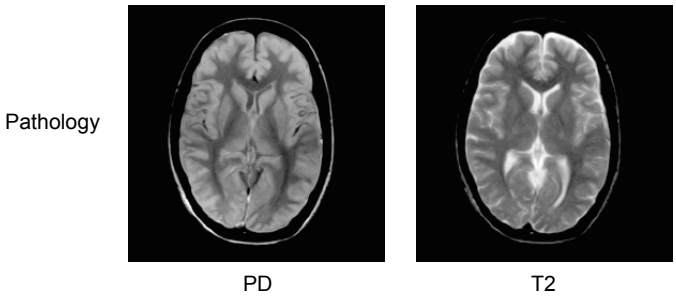
When To Use Which Contrast

With all these types of contrasts to choose from, one can ask which contrast to use in a particular situation. This is a valid question, but the answer is not so easy. Certain pathologies show better on PD weighted images than on T2 weighted images, while others show better on T1.

In general, though, we can stick to the following rule:



For a clear delineation of anatomical structures a T1 weighted or, even better, an IR sequence is the best choice (Page 43).



For pathology PD weighted or, preferably, T2 weighted contrast is used. The reason being that most pathology produces water (edema), which shows bright on T2 weighted images.

Figure 60

Another option would be to inject a MR sensitive contrast medium, such as Gadolinium-diethylenetriaminepentaacetic acid (Gd-DTPA or Gad for short; you got to be sick to think of names like that ☺), in which case you would scan a T1 weighted sequence because Gd-DTPA shortens the T1 relaxation time of tissue and shows, therefore, bright on a T1 weighted image. (Gd-DTPA does not show on T2 weighted images).

In practice almost always both T1 weighted (with and without Gd-DTPA) and T2 weighted sequences are scanned in one or more planes, to ensure optimum visualization of the pathology at hand.

Turbo Spin Echo Sequence

Although techniques such Multi-Slicing can be used to reduce the scan time, a regular SE T2 sequence can still take up to 12 minutes to acquire. During this time a practical problem arises: patient movement. It is very difficult to lie still for such a long time. And yet, that is necessary otherwise the image would be totally useless due to motion artifacts.
In order to reduce the scan time a very clever German chap called **Henning** came up with the Turbo-Spin-Echo (TSE) sequence (also known as Fast-Spin-Echo (FSE)).

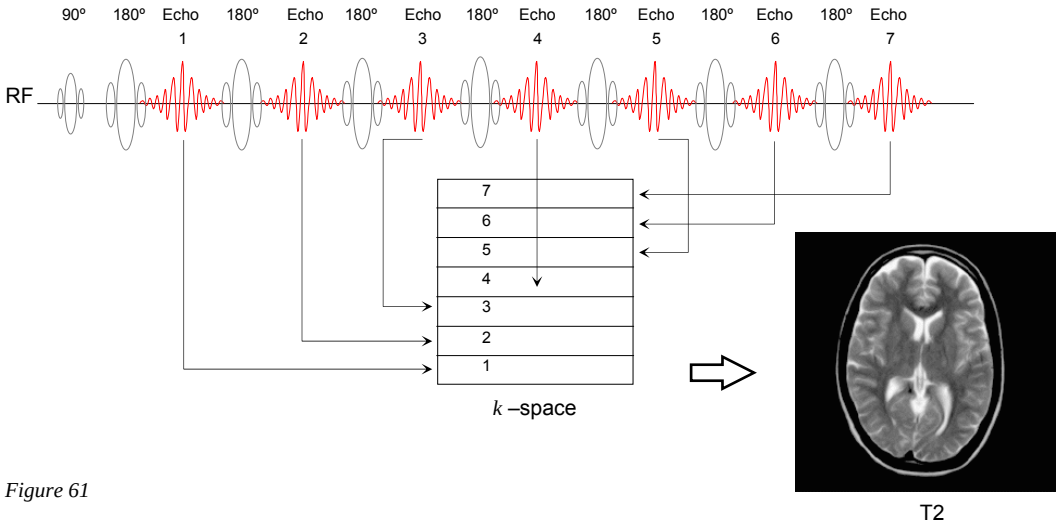


Figure 61

The TSE sequence also makes use of the multi-echo principle as shown in *Figure 61*. After the 90º pulse a series of seven 180º pulses are given. Each 180º pulse generates an echo.
k-Space is divided into 7 segments and each echo fills one line in each segment.
One, usually T2, image is reconstructed. The advantage of this technique is clear: a scan time reduction factor of 7. Compare these scan times:

Regular SE: TR 3000, TE 120, MX_{PE} 256 works out to 3000 x 256 = 12.8 minutes.

TSE: TR 3000, TE 120, MX_{PE} 256 and 7 echoes: (3000 x 256) ÷ 7 = 1.8 minutes.

You can understand that this type of sequence is very useful. Funnily enough many radiologists were not so keen to use this sequence. They were used to seeing a specific T2 contrast when scanning the brain. The FSE image has a mix of contrasts. *Figure 61* helps us to understand this. As we know signal and contrast information is stored in the centre of *k*-space. In our example you can see that the 4th echo, as well as parts of the 3rd and the 5th echo, is put in the centre of *k*-space. Because each echo is at a different time, each echo will carry different contrast information. The result is an image with a mix of contrasts.
Another negative aspect is that the image will show artifacts, which are specific for this type of sequence (see § Artifacts).
All in all in it took some time before TSE was widely accepted. There are radiologists, however, who still don't use TSE because of these drawbacks.

A series of echoes as used in TSE is called an Echo Train. One can choose how many echoes one would like to use. In our example we used an Echo Train Length (ETL) of 7 but an ETL of 212 is also possible.

It is also possible to create two images out of an Echo Train. All we need is two *k*-spaces. For example, with an ETL of 14, one can use the first 7 echoes for a PD image (1st *k*-space) and the

last 7 echoes for a T2 image (2nd *k*-space). This is called a Double-Echo TSE sequence or PD/T2 TSE sequence.

Fast Advanced Spine Echo or HASTE Sequence

We can take TSE a step further. In the Fast Advanced Spin Echo (FASE) sequence an ETL of 212 is used. This already results in ultra short scan times. On top of that Half Fourier Imaging is used. The combination of 212 echoes and HFI results in scan times, which are only a fraction of a regular SE sequence. *Figure 62* shows how it works.

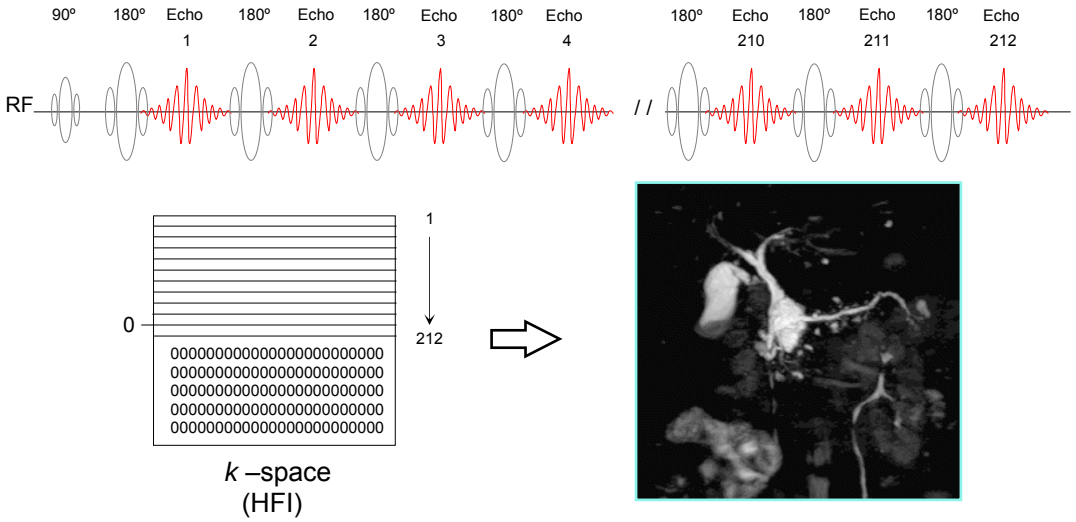


Figure 62

Each echo of this 212 ETL sequence fills one line in *k*-space. This fills *k*-space for slightly more than 50%. The rest of *k*-space is filled with zeroes (no data). The stunning bit is that you need only one repetition to create an image.
Notice that only the last (very late) echoes are put in the centre of *k*-space resulting in an image, which only shows free water (bile and water in the intestines as the image shown in *Figure 62)*. This kind of sequence is used for MR-Cholangio-Pancreatography (MRCP).

Gradient Echo Sequence

A second group of sequences are the Gradient Echo sequences. Also with this type of sequence an echo is rebuilt from the FID.

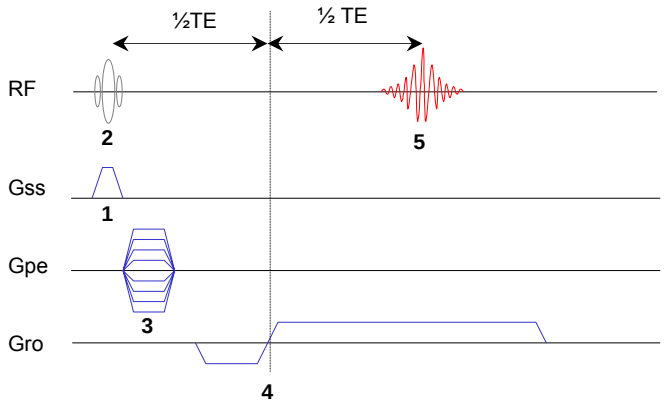


Figure 63

They differ from the Spin Echo sequence in the way the echo is formed. Where a Spin Echo sequence uses an 180º rephasing pulse to rephase the spins, the Gradient Echo sequence uses a gradient polarity reversal (*Figure 63*).

1. Slice selecting with G_{SS}.
2. Send excitation pulse.
3. Phase encoding.
4. Switch on G_{RO}. First negative polarity, and then change polarity to positive.
5. Signal sampling during G_{RO}.

Changing the polarity of the G_{RO} has the same effect as an 180º RF pulse. The advantage is that it can be done much faster than the 180º pulse. That makes this sequence useful when fast scans are needed. The disadvantage is that it does not correct for local magnetic field inhomogeneities, which translates into the presence of artifacts in the image.
The FA of the excitation pulse (this time written as α) can be anything in the range of 1º to 180º, although it depends very much on the contrast required. (Usually FA's between 1° and 90° are used).

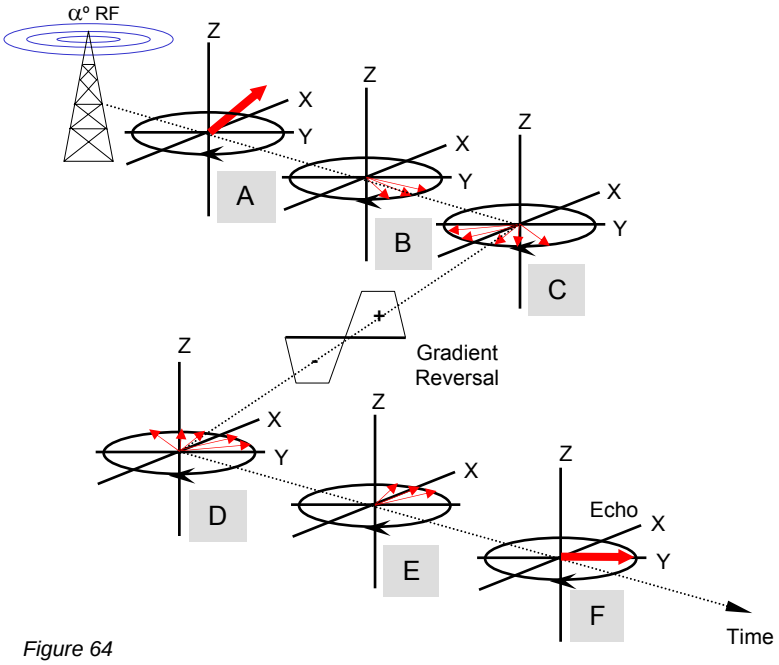


Figure 64

Figure 64 shows the sequence.

A. Start with a FA α. Depending on α, more or less magnetization is flipped into the X-Y plane.
B. The spins are dephasing…
C. And dephase a bit more until the polarity of the G_{RO} is changed after which…
D. The spins rephase…
E. And rephase a bit more until…
F. The spins are in-phase again.

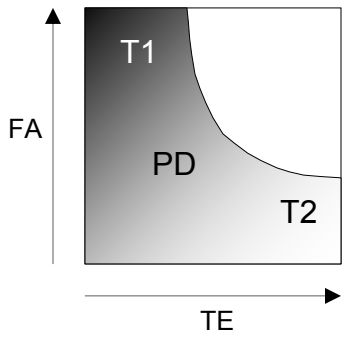


Figure 64A

The image contrast of GE sequences is determined mainly by FA and TE as shown in *Figure 64A*. A high FA and short TE gives T1 weighted contrast. A medium FA and short TE gives PD contrast. A low FA and long TE gives T2 weighted contrast. (See page 48 for an example of various FA's)

The appearance of GE images is quite different compared to SE images.
There are many variations on the GE sequence, which makes it a very versatile technique with its own range of applications.

Inversion Recovery Sequence

The last sequence technique is the Inversion Recovery (IR) sequence.

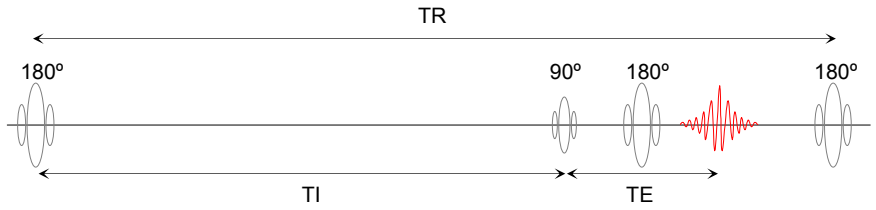


Figure 65

When you look at *Figure 65* you notice the difference. If you look at little closer you also see a striking resemblance with the SE sequence. In fact, an IR sequence is a SE sequence preceded by another 180º excitation pulse. The sequence dynamics are different from SE in that the first 180º excitation pulse flips the net magnetization to the $-M_Z$ axis. There is no magnetization in the X-Y plane as yet. After the 180º pulse there is only T1 recovery going on, because there is no component in the X-Y plane and, therefore, no T2 relaxation. The T1 relaxation process will take twice as long as when the net magnetization would have been flipped into the X-Y plane. T1 relaxation is allowed to happen for a certain time, known as the Inversion Time (TI). Then a normal SE sequence is applied to generate the image. Because an Inversion Recovery sequence usually has a fairly long TR (1500 ms.), and a short TE (10 ~30 ms.) the image contrast is almost entirely dependent on the Inversion Time (TI). The advantage of using this technique is that the T1 relaxation curves of the tissues are, so to speak, pulled away from one another in order to create higher T1 contrast differences.

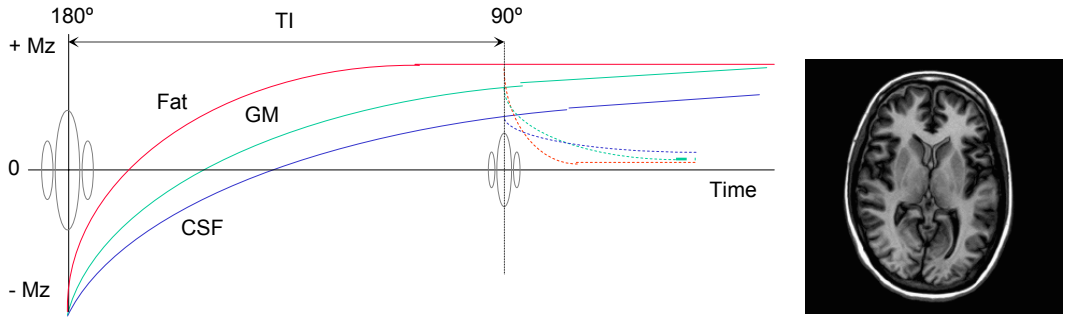


Figure 66

IR sequences are predominantly T1 weighted and the T1 contrast in the image is striking. For example, the T1 contrast difference of the various structures of the basal ganglia in the brain is pretty poor in a SE sequence. In an IR sequence, however, these structures are much better delineated (*Figure 66*).
Obviously one can wonder why IR sequences are not used commonly and the answer would be: scan time. The normal value for TR is in the range of 1500~2000 ms.. When we scan with a TR 2000, MX_{PE} 256 then the scan time would be 8.5 minutes. That's considerably longer than any SE T1 weighted sequence (2~3 minutes).
However, there is light at the end of the tunnel (and this time it's not from the oncoming train!). With the introduction of the TSE sequence it is possible to combine the two techniques. This reduces the scan time for IR sequences considerably, and is certainly worth using, if one accepts and appreciates the artifacts related to the TSE sequence.

In *Figure 67* some different values for TI are shown. Notice the effect TI has on the image contrast.

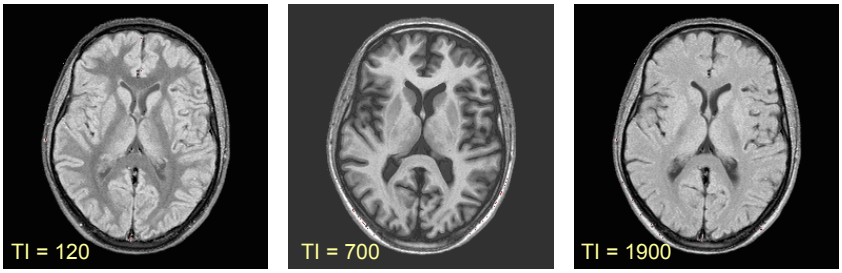


Figure 67

The huge impact the TI has on image contrast makes an IR sequence very versatile.
Normally a TI in the range of 400~700 ms. is used for 'normal' T1 weighted studies where good visualization of anatomy is important.
There are two 'special' cases concerning the choice of TI:

FLAIR (Fluid Attenuated Inversion Recovery) Sequence

An Inversion Time of 1900 ms. in combination of a long TE is used to study demyelinating diseases, such as Multiple Sclerosis. With this TI value, Multiple Sclerosis lights up like a light bulb. FLAIR is substantially more sensitive for demyelinating diseases than ordinary T2 weighted sequences.

STIR Sequence

When a TI of 160 ms. is used in a 1.5 Tesla system (90 ms. at 0,35 Tesla, 120 ms. at 0.5 Tesla, 140 ms. at 1.0 Tesla) something extraordinary happens. At 160 ms. the magnetization vector of fat tissue crosses the zero line. This means there is no vector pointing either towards $+M_Z$ or $-M_Z$. If one starts the SE part of the IR sequence at this particular time, there won't be any magnetization vector of fat available, which can be flipped into the X-Y plane; hence no signal from fat tissue is received.
This is a very effective way of suppressing signal from fat tissue, which is useful in those cases where the high signal from fat may obscure pathology such as bone bruises. This special case of IR sequence is called: Short TI Inversion Recovery (STIR).

Choosing The Right Sequence

Sequence Pro's And Con's

With all these sequences to choose from, such as SE, GE, IR or combinations of these, one might think it is virtually impossible to choose which sequence to use in a given situation. Luckily, it's not all that difficult. Most pathology can be detected with the good old SE sequence, either T1 or T2 weighted. But there are situation where one might want to use another sequence type.
In **table 2** the advantages and disadvantages of the various sequences are put together.

Table 2: Sequence Pro's and Con's

SEQUENCE	ADVANTAGE	DISADVANTAGE
(Turbo) Spin Echo	High signal Compensates for T2* effects "Real" T1 and T2 images	High RF deposit Long scan times Motion artifacts
Gradient Echo	Low RF deposits Short scan times Dynamic scan possibility	Low signal T2* related artifacts Motion artifacts
Inversion Recovery	High signal Real T1 images High T1 contrast Fat suppression	High RF deposit Very long scan times Limited number of slices Motion artifacts

Note that with all sequences motion artifacts are possible. However, nowadays it is possible to create sequences with such short scan times (while the patient is holding breath) that motion is virtually eliminated.

But it's not just these factors that determine the right choice. There are other issues such as Flow Compensation, Gradient and RF Spoiling, Pre-pulses, in- and out-phase imaging, fat suppression, cardiac- or peripheral and respiratory gating, Half Fourier Imaging to name but a few. It is beyond the concept of this story to discuss all these variations and applications. (Nearly got you worried there ☺).
It takes a bit of experience to choose the right sequence for a particular situation and experience only comes with lots of practice.

T1, T2 and PD Parameters

Table 3 shows parameter combinations to obtain a certain contrast weighting:

Table 3: Parameter Choices

SEQUENCE	TR	TE	TI	FA
SPIN ECHO				
T1	600	10~30		90
Proton Density	1000	10~30		90
T2	2000	80~250		90
Gradient Echo				
T1		2~14		60~90
Proton Density		2~14		30~60
T2		20~34		5~30
Inversion Recovery				
T1	2000	10~30	400~700	90
STIR	2000	10~30	80~120	90
FastFLAIR	5000	10~30	1800~2200	90

Remarks:
1. The TR in a GE sequence has little influence on image contrast.
2. The TI varies per field strength B_0. (1800 for 0.35T; 2200 for 1.5T)

Practical Physics II

Sequence Parameters

Of all the items we have discussed so far I believe that the next section about sequence parameters is the most important one. It is with these sequence parameters that you will have to work in practice and it is imperative that you understand in which way the parameters influence the final MR image. Once you understand the significance of each parameter you are fit to perform an MRI scan satisfactorily, and no sooner.

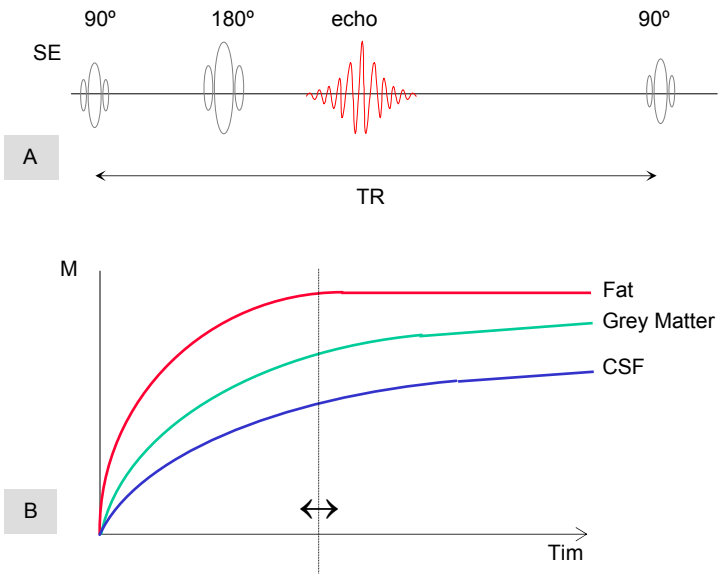
It's quite a chunk of information, and sometimes a bit confusing, but now we're going to piece things together. I'm convinced, once you understand this section, it will help tremendously in your daily routine.

The following parameters make up a MRI sequence:

TR	Repetition Time
TE	Echo Time
FA	Flip Angle
TI	Inversion Time
NA	Number Of Acquisitions
MX	Matrix
FOV	Field Of View
ST	Slice Thickness
SG	Slice Gap
PE	Phase Encoding
BW	Band Width

Repetition Time (TR)

The Repetition Time (TR) is the time between two excitation pulses (*Figure 68A*). In SE it's between two 90º pulses, in GE it's between two α pulses and in IR it's between two 180º pulses.

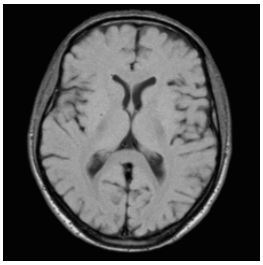


Increasing the TR (shifting the dotted line in *Figure 68B* towards the right) has the following effects on the image:

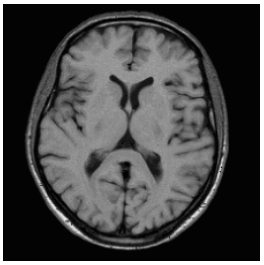
- Less image contrast.
  More time is allowed for T1 relaxation to take place; the difference in amplitudes of the magnetization vectors is smaller. Therefore…
- More PD contrast.
- More signal.
  There is more magnetization available for the next excitation.
- Increase of scan time.

Figure 68

C



TR 500 ms, TE 10



TR 1000 ms, TE 10

Example:

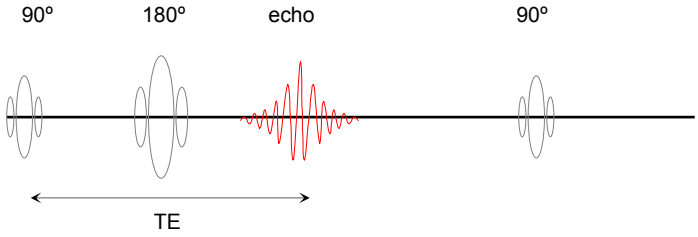
Figure 68C shows two images made with the same TE, but with different TR's. The image on the right has more PD contrast.

Figure 68C

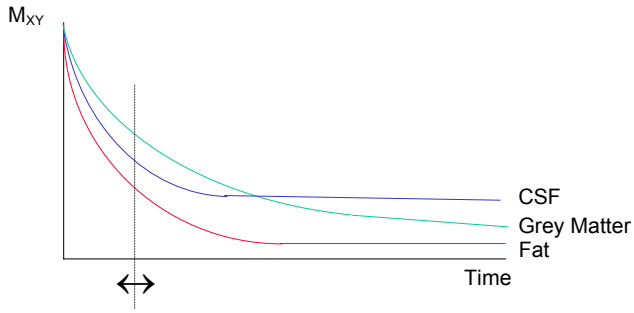
Echo Time (TE)

The Echo Time is the time between the excitation pulse and the echo (*Figure 69A*). It is an important parameter because the choice of TE influences image contrast dramatically in all types of sequences.

A



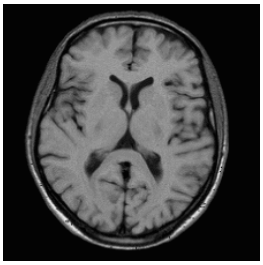
B



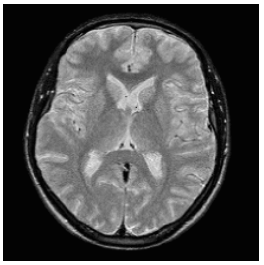
Increasing the TE (shifting the dotted line in *Figure 69B* towards the right) has these effects:

- More T2 contrast. An increase of TE allows for more dephasing.
- Less signal.
- Possible contrast swap. Notice that the relaxation curve of CSF in *Figure 69B* cross the one from gray matter. This means that with an early echo gray matter is brighter than CSF, while a late echo shows the opposite.

C



TR 2000 ms, TE 30



TR 2000 ms, TE 120

An example:

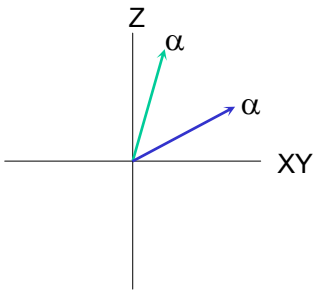
Figure 69C shows two images where the TR is kept the same, but the TE's are different. At 30 ms CSF is dark (PD contrast), while at 120 ms. CSF is bright (T2 contrast).

Figure 69

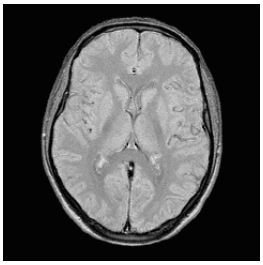
Flip Angle (FA)

The Flip Angle determines how much the net magnetization vector is rotated towards the X-Y plane (*Figure 70A*). In SE and IR sequences FA is most of the times 90º. In GE sequences, however, FA can have values in the range of 1º ~ 90º. In GE sequences the FA is, besides TE, responsible for the image contrast.

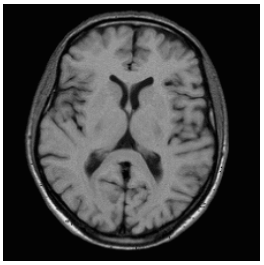
A



B



TR 150 ms, TE 10, FA 10



TR 150 ms, TE 10, FA 70

Figure 70

Increasing the FA (in a GE sequence) has these effects:

- More T1 contrast.
- More signal.
- Possible contrast swap.

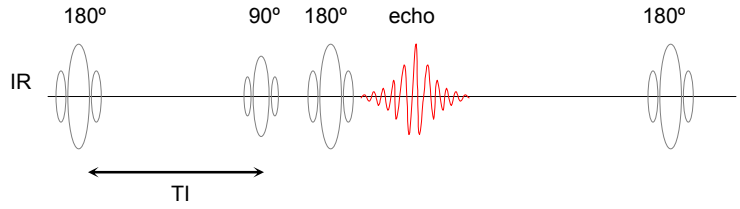
An example:

Figure 70B shows two images where TR and TE are kept the same, while changing the FA.
A low FA has more T2 weighting (CSF bright), and a high FA has more T1 weighting (CSF dark).

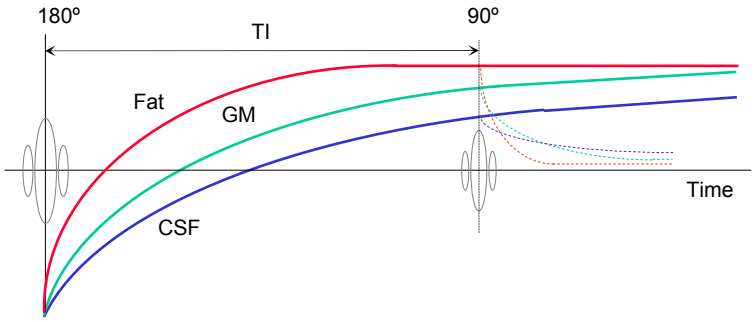
Inversion Time (TI)

The Inversion Time is the time between an 180º excitation pulse and the 90º-excitation pulse (*Figure 71A and 71B*).
TI is only used in IR type sequences and in a special kind of GE sequences (TurboGE).
TI has the highest impact on image contrast in IR sequences.

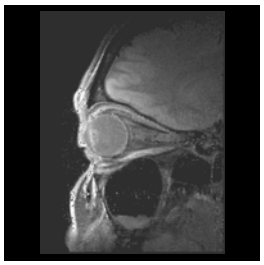
A



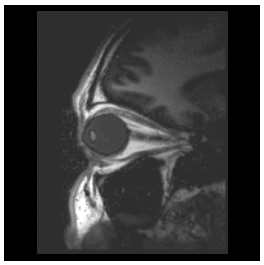
B



C



TR 2000 ms, TI 160 (STIR)



TR 2000 ms, TI 600

Figure 71

Increasing TI has these effects:

- T1 contrast change.
- More signal.

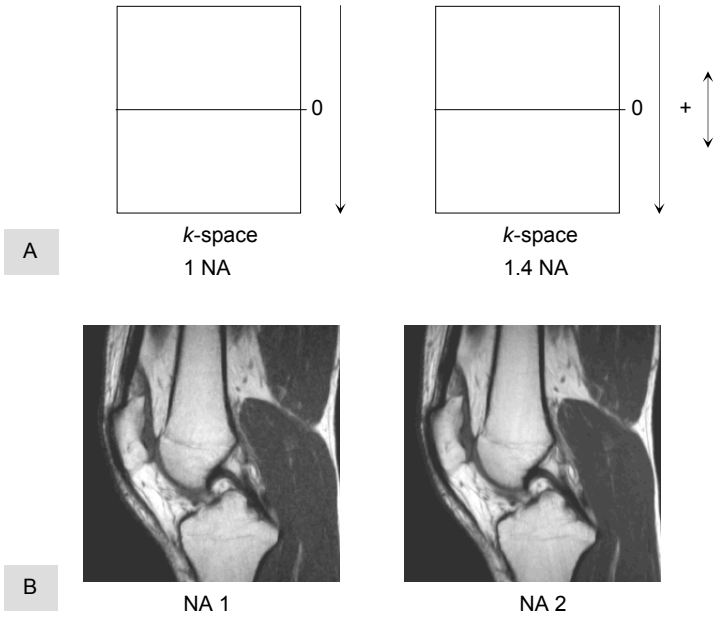
An example:

Figure 71C shows two images where TR is kept the same, while TI has different values.
The image on the left has a 'special' TI as seen in a STIR sequence. Here we see the retro orbital fat suppressed, which allows us to see the optic nerve in case of an MS plaque.
The image on the right is an ordinary IR image of the same eye.

Number Of Acquisitions (NA or NEX)

As has been explained earlier, an MRI image is reconstructed from *k*-space. To obtain a good image *k*-space needs to be filled completely from top to bottom. When we have filled *k*-space completely we say that we have done “1 acquisition”. In many cases the image quality in terms of Signal to Noise Ratio (SNR) is very poor. MRI offers the possibility to repeat the entire scan once or multiple times more. The extra signal that is acquired is then averaged resulting in a better SNR and therefore in a better image. The parameter Number Of Acquisitions (NA), also known as Number Of Excitations (NEX) or Signal Averaging determines how many times we repeat the entire scan.
There is, however, a catch: selecting NA=2 doubles the scan time, but the SNR is only increased by √2, which is 1.4 times!! In order to double the SNR, you’ll have to select NA=4, which quadruples the scan time. This is a serious parameter. Handle with care ☺.

Toshiba MRI scanners have a neat feature regarding NA. The systems offer the possibility to select fractional Number Of Acquisitions, such as 1.4 (*Figure 72A*). The thought behind this feature is: information regarding SNR and contrast are stored in the centre of *k*-space. To increase the SNR you only have to fill the lines around the centre of *k*-space again. There is no point filling the outside of *k*-space once more, because your image won’t get any sharper, no matter how many times you fill it. In the case of NA=1.4, 40% of the lines around the centre of *k*-space are filled once more. The advantage is that you only increase the scan time by a factor 1.4, instead of doubling the scan time as you would in the case of NA=2, while increasing the SNR considerably.



Increasing NA has these effects:

- More signal (√NA).
- “Less” noise.
- Fewer artifacts due to signal averaging. The higher NA, the better the image.
- Increase of scan time.

An example:

Figure 72B shows two images with different values for NA. The image on the right has more signal and fewer artifacts, but the scan time was twice as long.

Figure 72

Matrix (MX)

The (acquisition) Matrix determines the spatial resolution of our image. A matrix has two sides, MX_{PE} and MX_{RO}. The matrix size can usually be increased in steps of 32.
The acquisition matrix must not be confused with the display matrix. The display matrix can have two sizes 256 or 512. The acquisition matrix can have just about any size from 32 ~ 1024 with increments of 32. When an acquisition matrix of 192x256 is scanned, it will be reconstructed and displayed in 256x256 display matrix. Consequently, when an acquisition matrix of 192x320 is scanned, it will be reconstructed and displayed in a 512x512 display matrix.

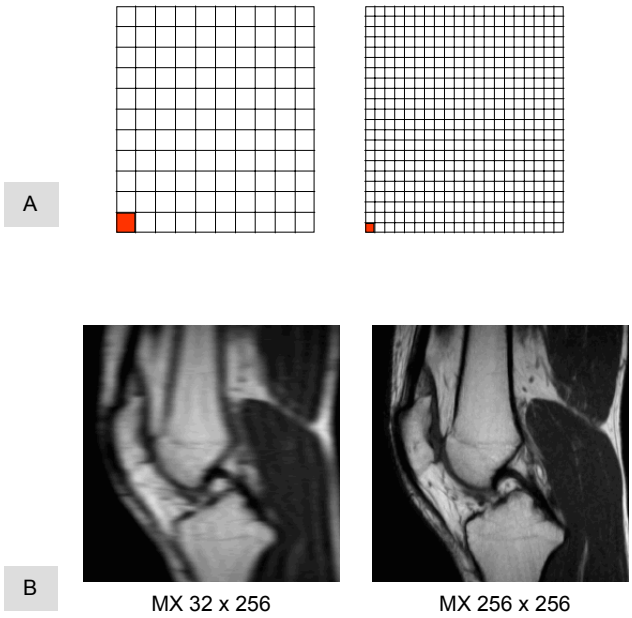


Figure 73

Increasing the acquisition matrix in any direction decreases the voxel size (*Figure 73A*), which has these effects:

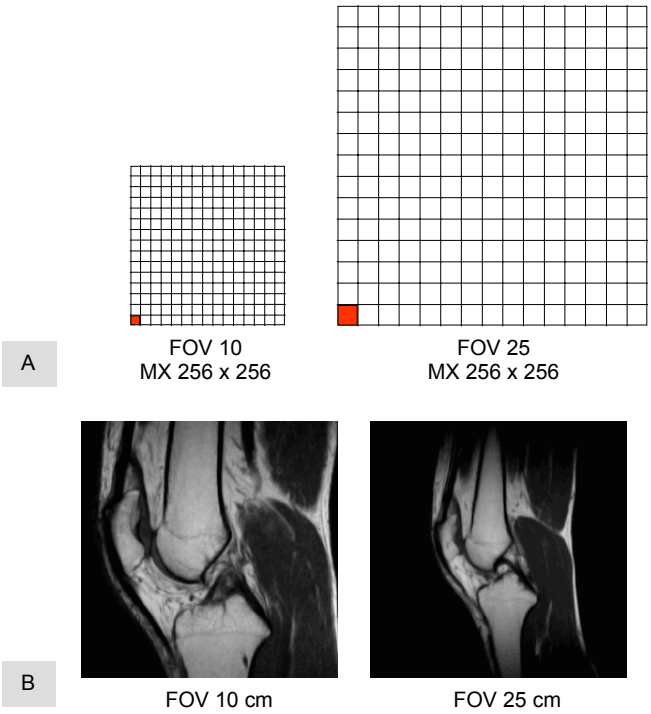
- Lower signal. A smaller voxel contains fewer protons, which can contribute to the signal/voxel.
- Higher spatial resolution.
- Increase of scan time. This only happens when MX_{PE} is increased (more lines are to be filled in *k*-space = more time). Increasing MX_{RO} has no effect on scan time.

An example:

Figure 73B shows two images with a different matrix size. (The first number indicates MX_{PE}). The image on the right has a higher matrix and is therefore sharper, but it took longer to acquire.

Field Of View (FOV)

The FOV determines how much of the patient we are going to see. A small FOV shows less than a big FOV does. This is pretty straightforward. Increasing the FOV size also increases the voxel size (*Figure 74A*).



Increasing the FOV has these effects:

- Increased signal. Increasing the voxel size also increases the number of protons, which can contribute to the signal/voxel. (SNR is increased by x^2).
- Lower spatial resolution.
- Increased viewing area.

An example:

Figure 74B shows two images with a different FOV. The image on the left with a FOV of 10 cm is sharper, it shows a smaller part of the body and it has lower SNR, than the image on the right.

Figure 74

Slice Thickness (ST)

The Slice Thickness influences the amount of signal, as well as the sharpness of an image. By changing ST from 10 mm to 5 mm we lose 50% signal. That is quite dramatic.

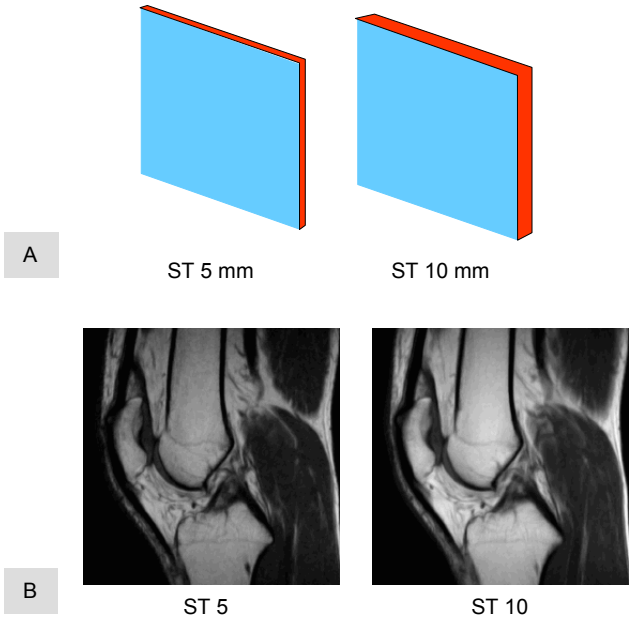


Figure 75

Increasing ST has these effects:

- Increased signal.
  The voxel size increases, so more protons can contribute to the SNR.
  $$SNR = \frac{ST_{modified}}{ST_{original}} * 100\%$$
- Lower resolution.
  This is straightforward.
- Increased "partial volume" effect.
  Partial volume effect occurs when an object, such as an adrenal gland is cut in half by a slice. If the signal would be high it would show on the image, but the size might be misinterpreted. Keep ST as thin as possible.
- Larger object coverage.
  20 slices of 5 mm cover 10 cm, while 20 slices of 10 mm cover 20 cm. (I love simple logic ☺).

An example:

Figure 75B shows two images with a different ST. The image on the right has increased signal, but is less sharp.

The Matrix (MX), the FOV and ST work together such as they all determine the voxel size (spatial resolution). All-important is to choose values for all three parameters, which result in enough SNR. For example:

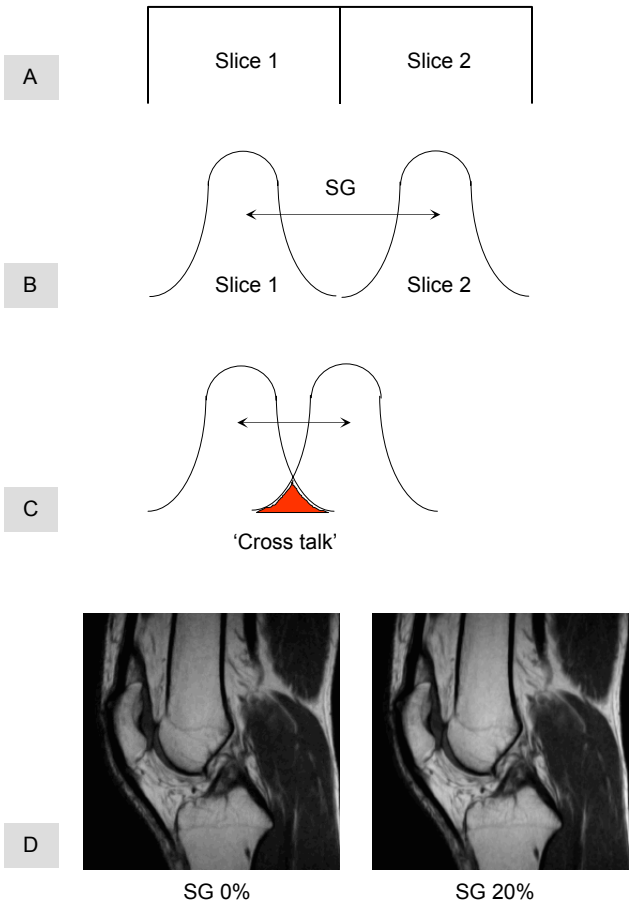
We scan a SE sequence with TR 500, MX 256x256, FOV 30x30, ST 6 and NA 1.
The voxel size is 30÷256 = 1.17mm x 1.17 mm x 6 mm.
The scan time is (500 x 256 x 1) ÷ 60000 ≈ **02:08** minutes.
The resulting SNR is normalized as 1.

To compare we make a scan with the same TR 500 and the same ST of 6 mm. We double MX to 512x512 and we reduce FOV by half to 15x15.
The voxel size is now 15÷512 = 0.29 mm x 0.29 mm x 6 mm, which is 4 times smaller.
However, (read this, this is fun) to maintain the same amount of signal, we would have to increase NA to 64, which increases scan time to **273:04** minutes (4 hours, 33 minutes and 4 seconds)!!!!
Anyone can see that this is not an option. Therefore, values for MX, FOV and slice thickness must be chosen such that there is enough signal to obtain a good image within a reasonable scan time. When selecting parameters, one makes a choice between SNR and contrast. An image with low SNR may still be useful as long as it shows the pathology.

That is not easy in the beginning, but after you have gained some experience you'll get a feel for it, and you will know exactly what parameter combinations are feasible with a particular sequence.

Slice Gap (SG)

The SG parameter describes the amount of space (in % of slice thickness) between slices.
In an ideal world the RF pulse, which creates the slice, would have a perfect slice profile as shown in *Figure 76A*. A perfect slice profile guarantees real contiguous slices without any space in between. In the real world slice profiles look more as shown in *Figure 76B*. There is an enormous gap in between the slices. In order to minimize this gap the slice profiles have to move closer to one another, which is possible, but it creates an overlapping area as shown in *Figure 76C*. When the slices overlap then an effect, known as "cross talk" occurs. The overlapping area contains signal from both slices. This signal is also seen in the resulting reconstructions.
In order to minimize cross talk effect we must have a gap between the slices. Usually a gap of 10% ~ 20% of the slice thickness is enough to minimize the cross talk effect.



Increase of SG has these effects:

- Less "cross talk".
- Increased coverage.

Figure 76D shows two images with different values for SG. Although it is extremely difficult to see, even on a monitor, the image on the left has cross talk effect.

Figure 76

There are other ways to scan without a slice gap. One way is to scan in "Interleave Mode", which acquires, for example, slices 1,3,5,7 first and after that slices 2,4,6,8. In Interleave Mode there is automatically a 100% gap, which eliminates cross talk altogether. (*The disadvantage of Interleave Mode is that the images may show signal intensity differences* ☹).

Phase Encoding (PE) Direction I

Although phase encoding is an excellent way to code the spins in order to find out where the signal originates, it causes also some serious problems.
One of the problems is an effect known as “Phase Wrap”. Phase Wrap occurs when the FOV is smaller than the object to be scanned.

If you choose a FOV as shown on the left in *Figure 78A*, your image will have an artifact as shown on the left in *Figure 78B*. The cause of this artifact is this:

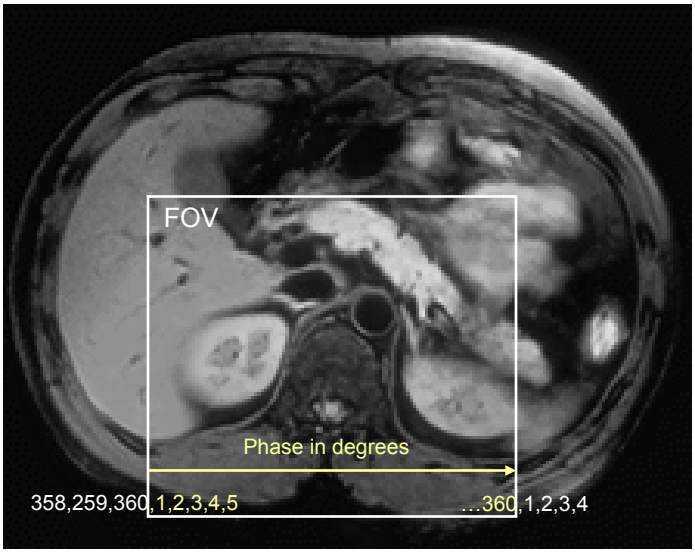


Figure 77

When phase encoding is performed, the FOV is divided into phase steps. In *Figure 77* phase 1 is at the left side *inside* the FOV. Phase 360 is on the right side *inside* the FOV.

The phase encoding gradient does not stop at the boundaries of the FOV. It continues encoding on the outsides of FOV. Phases can only be selected in the range from 1º ~ 360º. Once it has gone full circle, it will start again with 1 (as shown at the right hand side in *Figure 77*).
The system will not only receive signal from inside the specified FOV, but also from outside the FOV. It will receive a signal from *outside* the FOV (right hand side) with phase 1.
The computer, however, puts this signal *inside* the FOV *on the left side*, because that’s where it thinks phase 1 is located.
The same phenomenon happens on the other side of the FOV. So, we end up with an image as shown on the left in *Figure 78B*. The white band on the right hand side *inside* the FOV comes from the left hand side *outside* the FOV.
To counteract this artifact one simply turns on a feature known as “No-Wrap”, which doubles the FOV in the PE direction. An area, which is twice as big as the specified FOV is acquired, but ONLY the specified FOV is reconstructed, effectively eliminating the phase wrap artifacts (*Figure 78B right*).
There is a down side when turning on the No-Wrap option: it doubles the scan time.

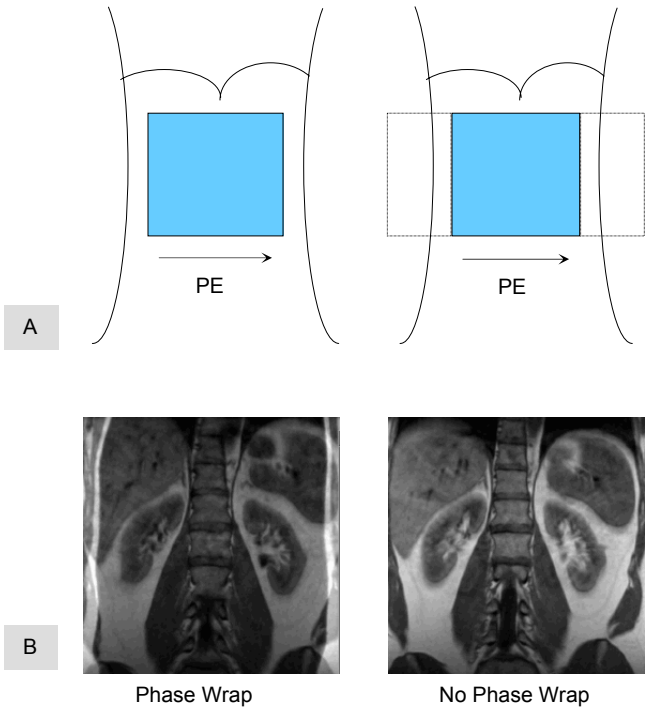


Figure 78

Phase Encoding (PE) Direction II

Another very important property of phase encoding is that it controls the direction in which motion artifacts are displayed. Motion artifacts are phase related. A motion artifact is created when a spin moves during the time between excitation and signal sampling (see also section about artifacts). A motion artifact is nothing more than mis-mapping of signal.

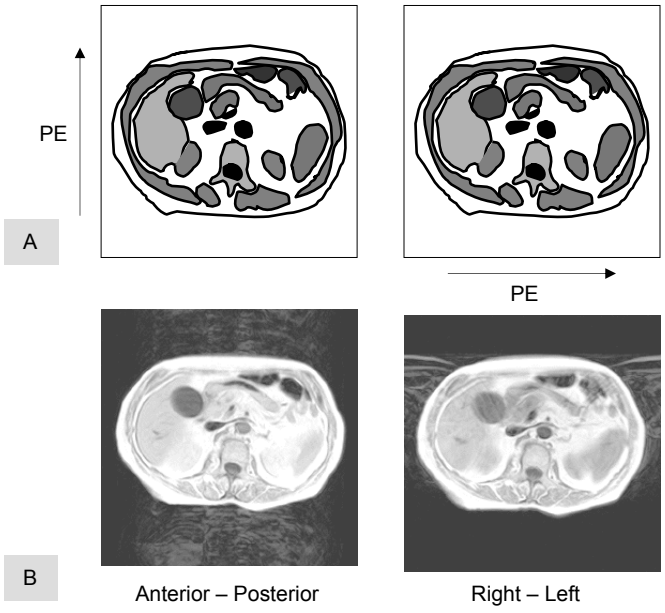


Figure 79

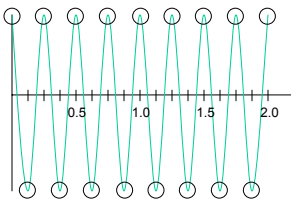
During set up a sequence you choose the direction of phase encoding. In *Figure 79A* an axial slice through the abdomen is scanned. You can choose the phase encoding direction to be in anterior-posterior or left-right direction. The resulting images in *Figure 79B* show motion artifacts due to breathing.

Choosing the phase encoding direction before scanning is very important. You must recognize what kind of motion, flow-, breathing- or pulsatile motion, you can expect to see and whether this is going to interfere with the area of interest. Many scans have to be repeated because of the wrong choice of phase direction. This choice is complicated by the fact that one has to take into consideration the possibility of phase wrap, as discussed before.

You can see more examples of motion artifacts in the section about artifacts.

Bandwidth (BW)

The signal we receive from the patient is continuous (analog). Computers, however, work with digital data. This means that the continuous signal has to be translated into a digital signal. The translation (digitization) is done by an Analog-Digital converter, which samples the amplitude of a signal at a certain rate. The **Nyquist Theorem** states that in order to represent a signal with a certain frequency accurately, you need a sampling rate that is twice as high.

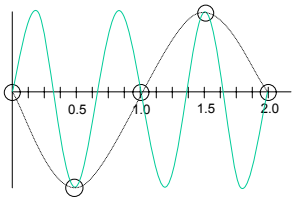


Period = 0.25 msec
Frequency = 4000 Hz

2 sample points per signal period

Sampling Rate = 1 point every 125 μsec = 8000 Hz

A



Sample Rate = 1 point every 500 μsec = 2000 Hz

Actual
Period = 0.6667 msec
Frequency = 1500 Hz

Apparent
Period = 2 msec
Frequency = -500 Hz

B

Figure 80

For example, if you want to digitize a frequency of 4000 Hz, you need a sampling rate of 8000 Hz (*Figure 80A*).

Receiver Bandwidth = Sample Rate = 1 ÷ Sample Time

Figure 80B shows a sine wave with a frequency of 1500 Hz. When we use a sampling rate of 2000 Hz, then the sine wave will be under sampled. The resulting reconstruction (dotted line) shows a –500 Hz sine wave. To adequately represent the 1500 Hz sine wave we would need a sampling rate of 3000 Hz.

Now we can put a few things together. The sampling bandwidth is in Hz. The gradient is defined in Hz/cm. The FOV is determined by the bandwidth divided by the gradient strength. The sampling bandwidth, in principle, defines the range of frequencies from one end of the FOV to the other.

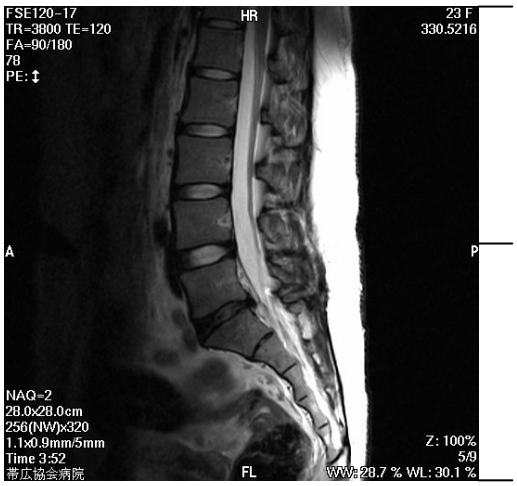


Figure 81

$$FOV = \frac{\text{Sample Bandwidth}}{\text{Gradient}_{RO}\ \text{Strength}}$$

Figure 81 shows an example of an image where a 28 KHz receiver bandwidth was used with a 1.0 KHz/cm read out gradient strength to get a FOV of 28 cm.

A smaller bandwidth means a slower rate of sampling; it takes longer to collect the same number of data points. Therefore, to get a smaller FOV you can either decrease bandwidth or increase the strength of G_{RO} (or a combination of both).

Remember though: taking a lower bandwidth will increase the TE, with more T2 decay (more T2 weighting).
Figure 81-2 shows the differences in SNR, TE and Chemical Shift in relation to low and high bandwidth.

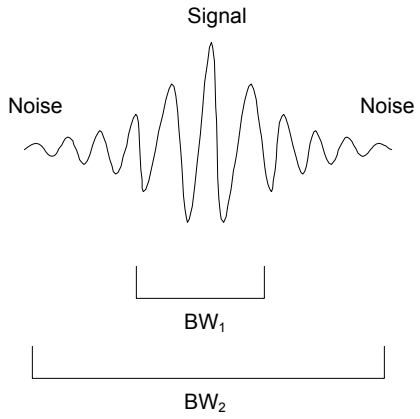


Figure 81-2

$SNR_{BW1} > SNR_{BW2}$

$TE_{BW1} > TE_{BW2}$

$\text{Chem. Shift}_{BW1} > \text{Chem. Shift}_{BW2}$

BW/2 = SNR + 40%

Practical Physics III

Image Artifacts

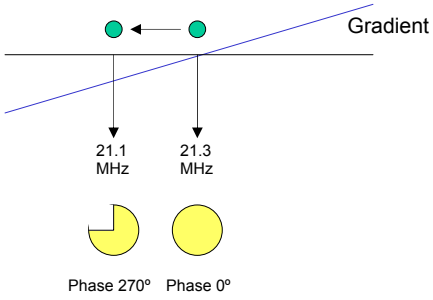
One can be sentimental about the quality of MR images, or one can rave about the excellent contrast resolution MR images can achieve, the fact is that any MR image is riddled with artifacts. It is important to recognize these artifacts and understand how they are created.

In this section I'll discuss the following artifacts:

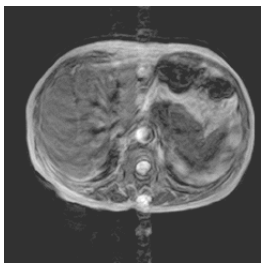
- Motion related artifacts
- Para-magnetic artifacts
- Phase Wrap artifacts
- Frequency artifacts
- Susceptibility artifacts
- Clipping artifact
- Chemical Shift Artifact
- Spike artifact
- "Zebra" artifact

Remember this: "Understanding artifacts enables you to read the MR image"

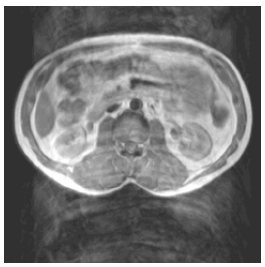
Motion Artifacts



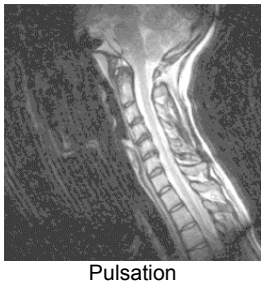
A



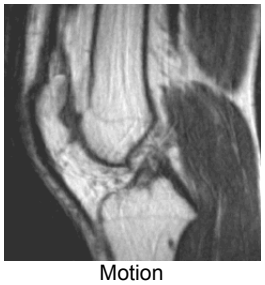
Pulsation



Respiration



Pulsation



Motion

B

Motion artifacts are caused by phase mis-mapping of the protons.

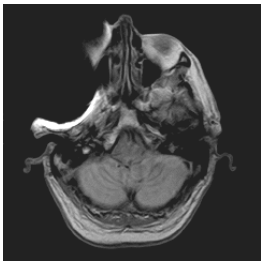
Because of the time-lapse between excitation and signal sampling the protons may have moved, due to respiration, pulsation or motion, through the gradient magnetic field, thus acquiring an additional phase shift. *Figure 82A* shows a spin with a frequency of 21.3 MHz and a 0º phase moving from right to left during the phase encoding gradient. During this movement the spin changes frequency and therefore also phase compared to its original position. When the image is reconstructed, the position of the signal is put in the wrong place in the image.
There are techniques, such as 'flow compensation' or 'cardiac triggering'; to minimize or eliminate motion related artifacts.
Motion artifacts are displayed in the phase encoding direction.

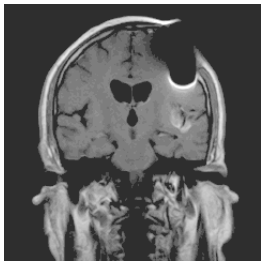
Figure 82B shows a few examples of motion artifacts.

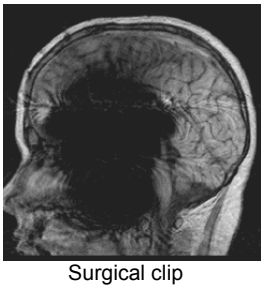
Figure 82

Para-Magnetic Artifacts

Para-magnetic artifacts are caused by metal (~ iron).
Metal deflects the magnetic field, thus changing the resonance frequency beyond the range, which is used in MRI. The protons will not react to the RF excitation pulse and will therefore not be displayed. What you may see are images as shown in *Figure 83*.


Metal splinter


Grenade shrapnel


Surgical clip

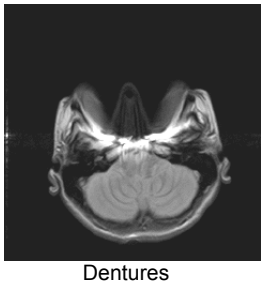

Dentures

Figure 83

Not all metals create such severe artifacts. It depends on the amount of iron. Very small pieces of iron, such as an iron splinter or surgical clips can cause havoc to the image.

Warning! Warning! Warning! Warning!

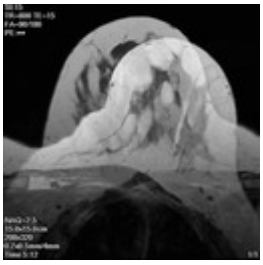
I cannot stress enough the importance of screening a patient for metal *before* the MRI examination. Apart from ruining the image, metal can also ruin the patient.

If there is any doubt whether an implant or other piece of metal is MR compatible, do NOT scan. Don't even get the patient close to the magnet (metal can kill, and that's no joke).

Aluminium and titanium produce much less severe artifacts. Patients with a titanium hip or knee implant can go into a MRI scanner without any problems.

Phase Wrap Artifacts

Phase wrap artifacts are caused by mis-mapping of phase.



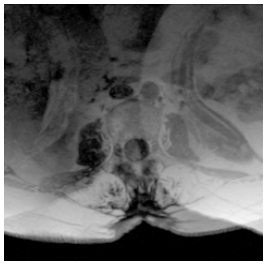


Figure 84

Phase wrap occurs when the Field Of View is smaller than the object. The part of the body outside the FOV will be 'wrapped around' into the image.
With the 'No Wrap' or 'Double Matrix' option switched on this artifact can be avoided, with a time penalty.

The image on the left in *Figure 84* shows the left breast being 'wrapped' into the image. The aim was to scan the right breast with a small FOV. The image on the right shows an axial lumbar spine. In both cases the operator forgot to switch on the No-Wrap option.

Frequency Artifacts

Frequency artifacts are caused by ‘dirty’ frequencies.

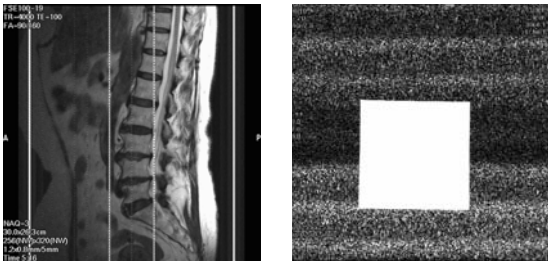


Figure 85

Faulty electronics, external transmitters, RF-cage leak, non-shielded equipment in the scanner room, metal in the patient, or when the door to the scanner room is left open can generate ‘dirty’ frequencies.
It usually requires an engineer to solve this kind of artifact, although, the door to the scanner room can be closed by non-qualified people as well ☺.
Frequency artifacts are displayed in the frequency encoding direction.

Susceptibility Artifacts

Susceptibility is the ability of substances to be magnetized, for example iron in blood.
Susceptibility artifacts are caused by local magnetic field inhomogeneity.

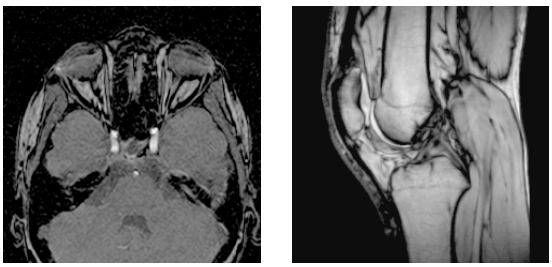


Figure 86

The different bonding properties of hydrogen protons in fat and muscle will cause local magnetic field inhomogeneity at the tissue boundaries. The resonance frequency at the boundaries has therefore changed and the protons involved will not be displayed. This can be seen in the image as a black line around the tissues, as if the image has been drawn with a pencil (*Figure 86*).

Clipping Artifact

Signal clipping or ‘over flow’ occurs when the receiver gain is set to high during the pre-scan.

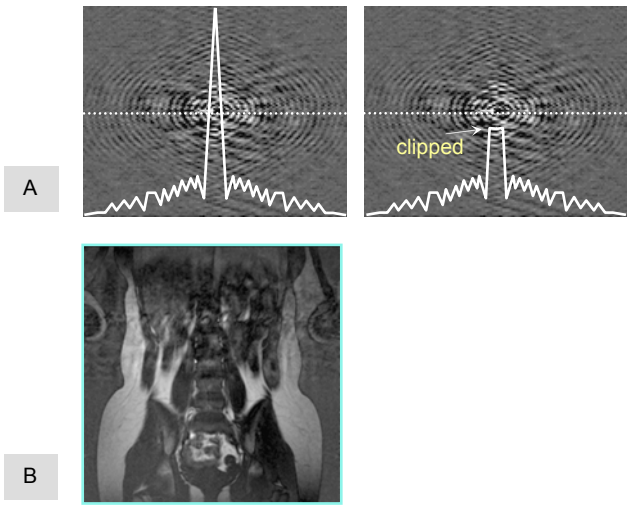


Figure 87

The maximum received signal is higher than the value set in the receiver gain (*Figure 87A left*).
The excess signal is clipped (*Figure 87A right*) and inverted and displayed in a different shade of gray (*Figure 87B*).

Chemical Shift Artifact

Chemical shift artifacts are caused by different resonance frequencies of hydrogen in lipids and hydrogen in water.

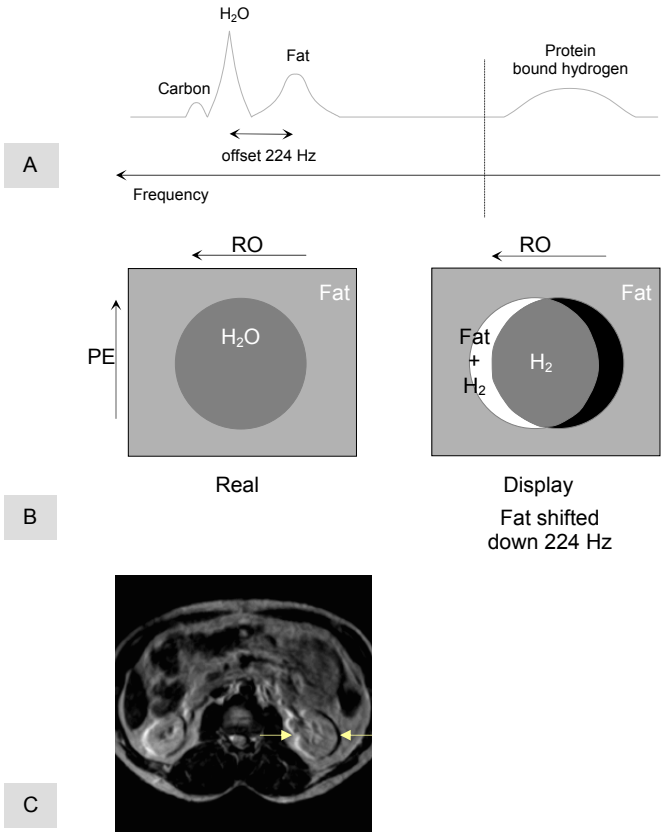


Figure 88

Figure 88A shows a frequency spectrum of a tissue sample.
At 1.5 Tesla the difference in resonance frequency between fat and water is 224 Hz (74 Hz at 0.5 Tesla, 52 Hz at 0.35 Tesla).

Figure 88B left shows a voxel containing both water and fat tissue, as can be found at the boundary between kidney and fat surrounding the kidney.
The Fourier Transform shifts the signal from fat a few pixels in the image, simply because it 'thinks' that's were the signal belongs (*Figure 88B right*).

Chemical shift happens in the frequency direction.

Figure 88C shows an example of chemical shift. The image shows clearly a white and a black border around the kidney. Fat is shifted towards the right, which means that the frequency-encoding gradient was in the right-left direction, while the phase-encoding gradient was in the anterior-posterior direction.

The chemical shift is related to the receiver bandwidth and FOV. The receiver bandwidth is defined as 1 / the time to sample one point. This represents the total range of frequencies from one end of the FOV to the other end. If we assume that a BW of 28 KHz (± 14 KHz) was used in *Figure 88* and a MX_{RO} of 256 then the frequency range per pixel is 28000 ÷ 256 = 109.375 Hz. The chemical shift at 1.5 Tesla is 224 Hz. Fat is shifted 224 ÷ 109.375 = 2 pixels.

Easy uh? (I consider this as advanced MRI physics, but I thought you might want to know ☺)

More on Chemical Shift can be found in the books by Westbrook and NessAiver mentioned in the § Recommended Reading

Spike Artifact

A spike artifact is caused by one ‘bad’ data point in *k*-space.

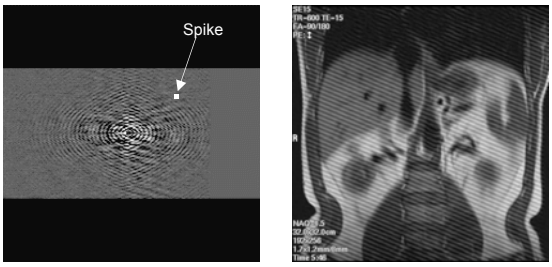


Figure 89

Figure 89 left shows one data point in *k*-space, which is out of the ordinary.

The resulting image (*Figure 89 right*) show diagonal lines throughout the image.

There is not a lot you can do to prevent this, apart from repeating the scan.

“Zebra” Artifact

The “Zebra” artifact may occur when the patient touches the coil, or as a result of phase wrap.

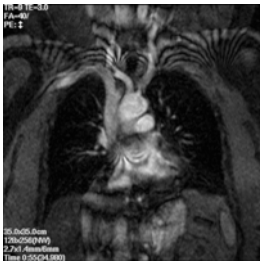


Figure 90

To avoid this problem you have to make sure that the patient is not touching the receive coil, or use No-Wrap option.

A final word about artifacts

There are many more different artifacts that can show up in a MR image. The ones that I’ve discussed here are the ones you might encounter frequently.
On the Internet site http://www1.stpaulshosp.bc.ca/stpaulsstuff/MRartifacts.html you can find many examples of other artifacts.

One word of caution: do not mistake an artifact for pathology. It has been seen that a motion artifact caused by a blood vessel was diagnosed as a tumour. Even when there is a shadow of a doubt, repeat the scan, either with a different technique or with different parameters.

Final Words

You must have a sore head by now and I don't blame you ☺.

There is one more thing, though, I would like to stress: **A synonym for MRI is COMPROMISE**.
It can't be said enough. In fact I'm going to say it again: A synonym for MRI is COMPROMISE.
I think this is the most important thing to remember from this story.
MRI is a balancing act for Signal to Noise. Whenever you change a parameter, you'll have to change another to counteract for the signal change, if you want to keep the SNR the same.
Many systems offer some sort of "SNR Indicator". When you change a parameter it shows how much signal you will gain or lose. It's a very nice tool, because sometimes it is not at all clear how much impact changing one parameter can have. If your system does not have a tool like that, then be very careful when changing parameters.

I hope that reading this story has helped you to understand the complex matter of MRI physics.
It only scratched the surface and you're by no means ready yet to build your own MR scanner, but at least you know what the abbreviation MRI means ☺.
I know it is not easy to get the total picture after reading the story once. Don't blame yourself. Read the story again, and again, if need be. Then open up a book from the reference list and try it again. One day there will be a moment that the penny drops and you will be rewarded with an immense feeling of happiness and you realize that MRI is even better than sliced bread.

These pages are the result of many years of teaching MR physics to radiographers and technicians who just started with MRI. They are by no means complete, but I hope it will encourage you to explore the exciting world of MRI a bit more. Take it in good time, though. Do not move on until you understand the basics. It is very easy to get lost in the myriad of subjects.

This is the right time to express my thanks to all those people who encouraged and supported me to put it all on paper; Aida Ferreira, one of my first students, who asked me to do it; my colleague Johan Roubroeks for test reading and, above all, Deirdre, my wife, for editing the story and who put up with the long hours.
Thanks all.

Have fun, ☺

Evert Blink

Appendix

Tissue Relaxation Times

Following **table 4** shows T1 and T2 relaxation times for various tissues (normal and pathological) at different magnetic field strengths.

Table 4: Tissue relaxation times

	T1 (ms) 0.25 T	T1 (ms) 0.5 T	T1 (ms) 1 T	T1 (ms) 1.5 T	SD (%)	T2 (ms)	SD (%)
BRAIN							
Gray matter	530	657	813	921	17	101	13
White matter	422	537	683	787	17	92	22
Tumours	667	802	963	1073	36	121	63
Meningioma	586	714	871	979	18	103	31
Glioma	845	887	931	959	35	111	33
Edema	667	806	973	1090	23	113	73
BONE							
Normal marrow	607	648	695	732	78	106	60
Osteosarcoma	740	811	888	973	28	85	30
BREAST							
Fibrotic tissue	409	547	732	868	18	49	16
Adipose tissue	190	214	241	259	28	84	36
Tumours	483	634	832	976	28	80	35
Carcinoma	451	595	785	923	25	94	48
Adenocarcinoma	490	686	959	1167	10	81	12
Fibroadenoma	508	715	989	1195	29	60	11
KIDNEY							
Normal tissue	417	496	589	652	27	58	24
Tumours	733	796	864	907	37	83	42
LIVER							
Normal tissue	250	325	423	493	22	43	14
Tumours	713	782	857	905	26	84	31
Hepatoma	621	769	951	1077	16	84	26
Chirrosis	328	367	410	438	21	45	
LUNG							
Normal tissue	488	599	735	829	19	79	29
Tumours	407	535	703	826	51	68	45
MUSCLE							
Normal tissue	409	547	732	868	18	47	13
Tumours	597	752	946	1083	32	87	40
Carcinoma	608	750	925	1046	16	82	73
Fibrosarcoma	831	896	967	1011	15	65	14
Rhabdomyosarcoma	664	827	1031	1173	27		
Edema	652	897	1235	1488	26	67	26
PANCREAS							
Normal tissue	302	371	455	513	25		
Tumours	718	942	1235	1448	15		
SPLEEN							
Normal tissue	431	543	683	782	19	62	27
Tumours						69	1

Acronyms

3D MP RAGE	3D Magnetization Prepared RAGE	Siemens	MR FE
AAS	Auto Active Shimming	Toshiba	MR special
ACE	Accelerated Contrast Enhancement (Keyhole Imaging)		MR
ACR	American College of Radiology		General
ACS	Advanced Clinical System	Philips	MR
ADC	Analog-to-Digital Converter		General
AFP	Adiabatic Fast Passage		MR
AIE	ACCESS Integrated Electronics	Toshiba	MR
AIP	Advanced Image Processing (ACCESS LPT)	Toshiba	MR
BEC	Both Echo CFAST	Toshiba	MR Special
BEST	Blood vessel Enhancement by Selective Suppression Technique	Toshiba	MRA
BFAST	Blood Flow Artifact Suppression Technique	Toshiba	MR Special
CCU	Coronary Care Unit (in Hospital)		General
CE-FAST	Contrast Enhanced FAST	Picker	MR FE
CFAST	CSF Flow Artifact Suppression Technique	Toshiba	MR FE
ChemSat	Chemical Saturation	GE	MR Special
CHESS	CHEmical Specific Saturation	GE/Sie	MR Special
CIDNP	Chemically Induced Dynamic Nuclear Polarization		MR
CMSMP	Cycled Multi Slice Multi Phase SE Acquisition	Philips	MR
CNR	Contrast to Noise Ratio		General
CNR	Contrast-to-Noise Ratio		General
COPE	Centrally Ordered Phase Encoding	Picker	MR Special
COSY	Correlation Spectroscopy		MRS
CP-COIL	Circular Polarized coil (similar to QD-coil)		General
CPMG	Carr-Purcell-Meiboom-Gill sequence		MR SE
CSA	Chemical Shift Anisotropy		MRS
CSF	CerebroSpinal Fluid		General
CSFSE	Contiguous Slice FSE	UCLA/GE	MR SE
CSI	Chemical Shify Imaging	ACR	MRS
CSMEMP	Contiguous Slice MEMP	GE	MR SE
CSPAMM	Complementary-SPAMM	Philips	MR
CYCLOPS	CYCLically Ordered Phase Sequence		MR
DAC	Digital-to-Analog Converter		General
DANTE	Special pulse????		MRS
DAVE	Dual Acqisition with Variable Echotimes	Philips	MR SE
DE FGR	Driven Equilibrium FGR	GE	MR FE
DEFAISE	Dual Echo Fast Acquisition Interleaved SE	UCLA	MR SE
DIET	Dual Interval Echo Train	Toshiba	MR Special
DMIA	Dig. Modality Interface Adaptor (MERGE-box)	Philips	DI
E-SHORT	SS-GRE with SE sampling (non-spoiled	Elscint	MR FE
ECG	Electro-CardioGram		General
EIS	External Interference Shield	Siemens	MR
EPI	Echo Planar Imaging		MR
ESDI	Extended Small Device Interface		General
ESMD	Enhanced Storage Module Device		General
F-SHORT	FID based SHORT	Elscint	MR FE
FACE	FID Acquired Echo		MR FE
FAISE	Fast Acquisition Interleaved SE	Harvard	MR SE
FAME	Fast Acquisition Multiple Excitation		MR
FASE	Fast Advanced SE	Toshiba	MR Special
FASE	FAst SE	Yale	MR SE
FAST	Fourier Acquired Steady-state Technique	Picker	MR FE
FATSAT	FAST SATuration	GE	MR Special
FDDI	Fiber Distributed Data Interface		DI
FE	Field Echo	Pic/Tos	MR FE

FEDIF	Field Echo with echo time set for water/fat signals in opposition	Picker	MR FE
FEER	fast Field Echo with Even echo Rephasing		MR FE
FESUM	Field Echo with echo time set for water/fat signals in phase	Picker	MR FE
FFE	Fast Field Echo (Gradient Echo sequence)	Philips	MR FE
FFT	Fast Fourier Transform		General
FGR	Fast GRass	GE	MR FE
FID	Free Induction Decay	ACR	MR
FIS	Free Induction Signal		MR
FISP	Fast Imaging with Steady Precession	Siemens	MR FE
FLAG	Flow Adjustable Gradient sequence	GE	MR Special
FLAIR	Fluid Attenuated IR		MR
FLAK	FLow Artifact Killer		MR Special
FLARE	FLip ? Acquisition with Relaxation Enhancement	Hennig	MR SE
FLASH	Fast Low Angle SHot	Siemens	MR FE
FLOWCOMP	FLOW COMPensation	GE	MR Special
FOI	Field of Interest		General
FOV	Field of View		General
FREEZE	respiratory selection of phase-encoding steps	Elscint	MR Special
FRODO	Flow and Respiratory artifact Obliteration with Dir		MR Special
FSE	Fast Spin Echo	GE	MR SE
FSPGR	Fast SPGR	GE	MR FE
GE	Gradient Echo	ACR	MR FE
GFE	Gradient Field Echo	Hitachi	MR FE
GFEC	Gradient Field Echo with Contrast	Hitachi	MR FE
GMN	Gradient Moment Nulling		MR Special
GMR	Gradient Moment/Motion Reduction/Rephasing	Siemens	MR Special
GRASE	GRAdient and Spin Echo	Harvard	MR SE
GRASS	Gradient Recalled Acquisition in Steady State	GE	MR FE
GRE	Gradient Recalled Echo	GE/Sie	MR FE
GREC	GRadient Echo with Contrast	Hitachi	MR FE
GRECO	Gradient Recalled EChO	Resonex	MR FE
HASE	High Angle Spin Echo (see MSE)	GE	MR SE
HASTE	Half Fourier Single shot Turbo spin Echo	Siemens	MR Special
HEAT	Hadamard Encoded Acquisition Technique	Elscint	MR
HEPI	Hybrid Echo Planar Imaging	Toshiba	MR Special
ICU	Intensive Care Unit (in Hospital)		General
ICV	Intra-Cranial Vessel		General
IIP	Interactive Image Processor	Toshiba	MR
ILOPS	Image LOcalized Phosphorus Spectroscopy (MRSI)		MRS
ILS	Image Localized Spectroscopy		MRS
IR	Inversion Recovery		MR
IRFGR	IR Fast GRass	GE	MR FE
IRTSE	IR voor TSE	Philips	MR SE
ISCE	Inclined Slab for Contrast Enhancement	Toshiba	MRA
ISDN	Integrated Services Digital Network	PTT	DI
ISIS	Image Selective In-vivo Spectroscopy	Philips	MRS
ISO	International Standardization Office		General
JIPS	Joint Image Processing System (Gyroview)	Philips	DI
LASE	Low Angle SE (see MSE)		MR SE
LPT	Laminated Polar Tip (ACCESS LPT) or Low field Permanent magnet Technology	Toshiba	MR
MAST	Motion Artifact Suppression Technique	Picker	MR Special
MBW	Matched BandWidth technique	Toshiba	MR
MEG	Magneto-Encephalo-Graphy (SQUID)		General
MEM-MO-RE	Multi Echo Multi MOment REfocussing		MR Special
MEMP	Multi Echo Multi Planar	GE	MR SE
MIP	Maximum Intensity Projection	ACR	MRA
MOD	Magneto-Optical Disc (erasable)		General

MOTSA	Multiple Overlapping Thin Slab Acquisition	GE	MR
MPGR	Multi Planar Gradient Recalled	GE	MR FE
MPRAGE	Multi Planar RAGE		MR FE
MR	Magnetic Resonance		MR
MRA	MR Angiography		MR
MRI	MR Imaging	ACR	MR
MRM	MR Mammography		MR
MRS	MR Spectroscopy		MRS
MRSI	MR Spectroscopy Imaging		MRS
MRV	MR Video loop (=movie display)	Siemens	MR
MSE	Modified Spin Echo	Philips	MR SE
MSMP	Multi Slice Multi Phase		MR
MSOFT	Multi Slice Off-resonance Fat Suppression Technique	Toshiba	MR Special
MSS	Multi Single Slice Technique		MR
MTC	Magnetization Transfer Contrast	ACR	MR
MTMRI	Motion Triggered MRI	Kiel	MR
MTS	Magnetization Transfer Suppression		General
NAQ	Number of Acquisitions	Toshiba	MR
NEMA	National Electronic Manufacturers Association		General
NEX	Number of EXcitations = NSA	GE	MR
NMR	Nuclear Magnetic Resonance		General
NOE	Nuclear Overhauser Effect	ACR	MRS
NSA	Number of Signals Averaged	ACR	MR
OD	Optical Disk (non-erasable)		General
OSI	Open Systems Interconnection (ISO)		DI
PACS	Picture Archiving ansd Communication System		DI
PASTA	Polarity Altered Spectral and Spatial Selective Acquisition Technique	Toshiba	MR Special
PCA	Phase Contrast Angiography		MRA
PCR	Philips Computed Radiology	Philips	DI
PD	Proton Density		General
PEAR	Phase Encoded Artifact Reduction technique	Philips	MR Special
PFI	Partial Flip Imaging	Toshiba	MR FE
PIETIR	Prolonged Inversion Echo Time IR	Picker	MR
POMP	Phase Offset Multi Planar	GE	MR
PPM	Parts Per Million		General
PREP	Partial Rf Echo Planar	GE	MR FE
PRESS	Point Resolved Spectroscopy		MRS
PRIME	Proton Regional Imaging of MEtabolites technique	Philips	MRS
PRIME	PRe Inversion Multi Echo	Picker	MR
PSIF	mirrored FISP		MR FE
PURE	Precise and Ultimate Radiowave Electronics (RF)		General
QD-COIL	Quadrature Detection coil (similar to CP-COIL)		General
QUEST	QUick Echo Split imaging Technique	Siemens	MR FE
RACE	Real-time ACquisition and Evaluation	Siemens	MRA
RAGE	RApid Gradient Echo		MR FE
RAMFAST	Reduced Acquisition Matrix FAST techniques	Picker	MR FE
RARE	Rapid Acquisition Relaxation Enhanced SE	Hennig	MR SE
RASE	Rapid Acquisition SE	Philips	MR SE
RAVE	Reconstruction Algorithm for Voxel Enhancement	Toshiba	MR Special
RCOMPL/H	Resp compensated l/h sort (=PEAR)	GE	MR Special
RESCOMP	RESpiratory COMPensation	GE	MR Special
REST	REgional Saturation Technique	Philips	MR Special
RF-FAST	RF-spoiled FAST	Picker	MR FE
RFOV	Rectangular FOV		General
RICE	Refocusing Irregular pulse for Contrast Enhancement	Toshiba	MR Special
RISC	Reduced Instruction-Set Computer		General
RISE	Rapid Inversion SE	Picker	MR SE

ROAST	Resonant Offset Averaging in STeady STate	Siemens	MR FE
ROC	Receiver Operation Characteristics (im. Quality)		General
RODEO	ROtating Delivery of Excitation Off-resonance		MRA
ROPE	Respiratory Ordering of Phase Encoding	Picker	MR Special
RS	Rapid Scan	Hitachi	MR
RSPE	Respiratory-Sorted Phase Encoding		MR Special
SAR	Specific Absorption Rate	ACR	MR
SASI	Suggart Assembly System Interface		DI
SAT	SATuration	GE	MR Special
SCSI	Small Computer System Interface		DI
SE	Spin Echo sequence	ACR	MR SE
SECSY	SE-Correlated Spectroscopy		MRS
SETSE	Shifted Echo TSE	Philips	MR SE
SFP	Steady state Free Precession	ACR	MR FE
SGI	Silicon Graphics Incorporated		General
SHORT	SHORT repetition techniques	Elscint	MR FE
SIMP	Scaled Integration of Maximum Pixel		MRA
SIST	Summed Imaging after Soft Tresholding		MRA
SMART	Sampled Motion Artifact Reduction Technique	Philips	MR Special
SMASH	Short Minimum Angle SHot	Shimadzu	MR FE
SNR	Signal to Noise Ratio		General
SORS	Slice Selective Off Resonance Sync Pulse	Toshiba	MRA
SPAMM	SPAtial Modulation of Magnetization (tagging)		MR
SPARS	SPAtially Resolved Spectroscopy	Philips	MRS
SPGR	Spoiled GRass	GE	MR FE
SPIR	Spectral Presaturation with Inversion Recovery	Philips	MR Special
SR	Saturation Recovery sequence		MR
SSFP	Steady State Free Precession	Siemens	MR FE
SSMP	Single Slice Multi Phase		MR
STAGE:T1W	Small Tip Angle Gradient Echo: T1W	Shimadzu	MR FE
STEAM	STimulated Echo Acquisition Mode	Siemens	MRS
STERF	Steady-state TEchnique with Refocussed FID	GE/Shim	MR FE
STIR	Short TI IR	Philips	MR Special
T1	Longitudinal Relaxation time (spin-lattice)		MR
T1-FFE	(F)FE with enhanced T1 contrast	Philips	MR FE
T1-TFE	TFE with enhanced T1 contrast	Philips	MR FE
T2	Transverse Relaxation time (spin-spin)		MR
T2-FFE	FFE with enhanced T2 contrast	Shimadzu	MR FE
T2-TFE	TFE with enhanced T2 contrast	Philips	MR
TE	Echo Time		MR
TFE	Turbo Field Echo	Philips	MR FE
TFL	Turbo Flash	Siemens	MR FE
TGSE	Turbo Gradient and Spin Echo (=GRASE)	Siemens	MR SE
TI	Inversion Time		MR
TOF	Time Of Flight		MRA
TONE	Tilted Optimized Non-saturated Excitation	Siemens	MRA
TR	Repetition Time		MR
TRAP	Traced Ray Array Processing	GE	MRA
TSE	Turbo SE	Phi/Sie	MR SE
UFGRASS	Ultra Fast GRASS (TFE)	GE	MR FE
UFGRE	Ultra Fast GRE	GE	MR FE
UPS	Uninterruptable Power Supply		General
VE	Variable Echo	GE	MR
VEMP	Variable Echo Multi Planar	GE	MR
WCHASE	Water Chemical Selective Excitation	Toshiba	MR Special
WPAST	WraP-Around Suppression Technique		MR

Recommended Reading

Physics

TITLE	AUTHOR	PUBLISHER	ISBN
* MRI made easy (…well almost)	H.H. Schild	Shering	3-921817-2
Magnetic Resonance Imaging Physical and Biological Principles	S.C. Bushong	Mosby	0-8016-1820-7
* MRI in practice	C. Westbrook/C. Kaut	Blackwell	0-632-03587-0
* All, you really need to know, about MRI physics	M. NessAiver	Simply Physics	0-9660982-0-X
Fast Scan Magnetic Resonance Principles and Applications	F.W. Wehrli	Raven Press	0-88167-746-9
* Questions and Answers in MRI	A.D. Elster	Mosby	0-8016-7767-X
Magnetic Resonance Imaging Volume 1	D. Stark / W. Bradley	Mosby	0-8016-4930-7
Pocket Guide to MR procedures And metallic implants.	F. Shellock	Lipincott-Raven	0-397-51751-3

Clinical

TITLE	AUTHOR	PUBLISHER	ISBN
Handbook of MRI technique	C. Westbrook	Blackwell	0-632-03884-5
Magnetic Resonance in Medicine	P.A. Rinck	Blackwell	0-632-03781-4
The MRI manual	R.B. Lufkin	Mosby	0-8151-5593-X
Magnetic Resonance Imaging Volume 2	D. Stark / W. Bradley	Mosby	0-8016-4930-7

MRI On The Internet

There are a zillion websites dealing with MRI. The ones mentioned here contain links to other sites. You can spend the rest of your life reading them, might you choose to do so ☺.

Physics

INTERNET SITE	ADDRESS
Joseph P. Hornak, Ph.D.	http://www.cis.rit.edu/htbooks/mri
The Adelaide MRI Website (portal)	http://www.users.on.net/vision/
MRI Safety	http://www.mrisafety.com

** An absolute "must" read.*

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About the author

Was trained as a radiographer at the Rooms Katholiek Ziekenhuis, Groningen, The Netherlands, where he worked in the field of general radiography for three years. Following this, he completed the training for radiotherapy technician at the Rotterdams Radio-Therapeutisch Instituut (RRTI), Rotterdam, The Netherlands and worked there for nearly three years. During this period he also worked with one of the first CT-scanners in Rotterdam. He then moved back to Groningen where he worked with CT and MRI for six years in the University Hospital (AZG). During this period he was also trained in Ultra-Sound. After a three-year period working with MRI in the Military Hospital in Riyadh, Saudi Arabia, he started to work for Toshiba Medical Systems Europe, Zoetermeer, The Netherlands in 1991 as an Application Specialist MRI.

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