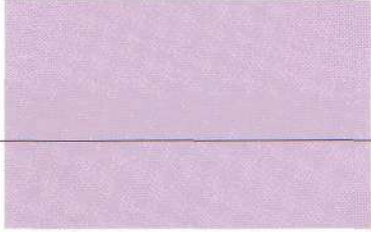


SCHERING

MRI

made easy

(...well almost)



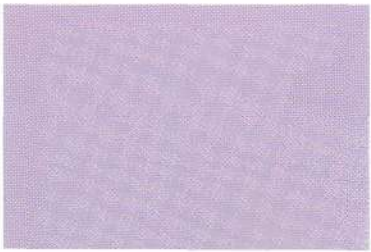
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Preface

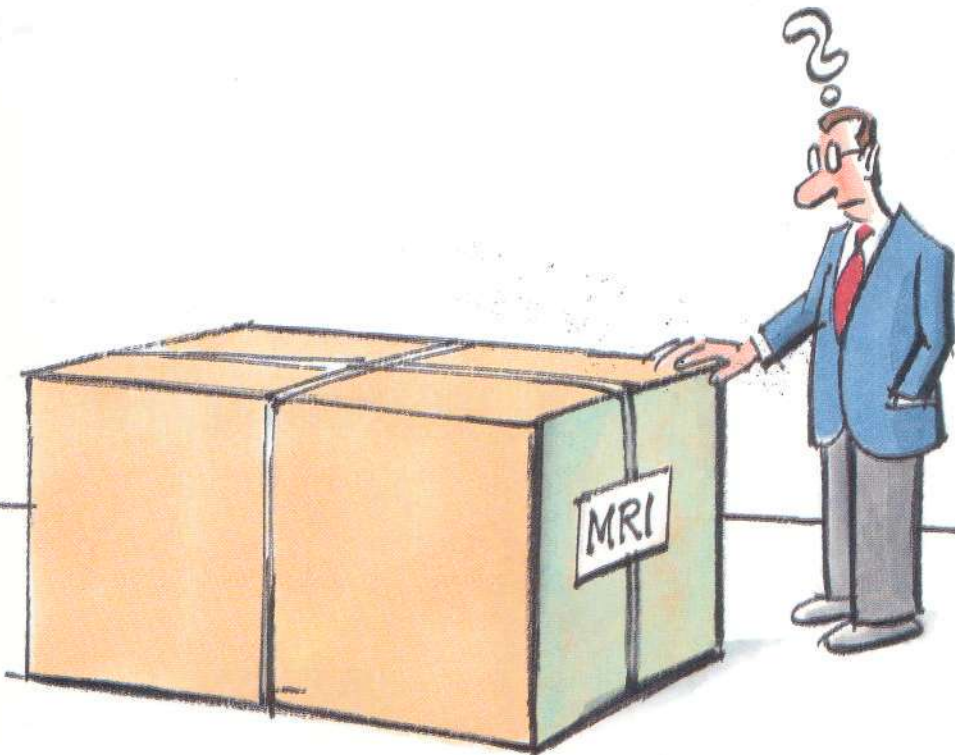
This book is dedicated

- to anyone, who tries to teach medicine instead of just reporting medical facts (like my anatomy teacher, Prof. Dr. R. Bock, who is a master of this art).

- and to anyone, whose stumbling feet find the MRI path difficult (The book was written in the hope rather than the belief that they may find some help from it).

(modified from Alastair G. Smith, Surgeons Hall, Edinburgh, October 1939)

H. H. Schild



About this book

This book was written as an introduction to magnetic resonance imaging (MRI). It is dedicated to anyone, who would like to know something about MRI without having to study physics for years. If this applies to you, then read this text from front to back, though not at one sitting. While the subject matter is extremely complex, it is not by any means beyond comprehension. It does however, require some concentration and consideration. I have therefore on occasion suggested that you set the book down and take a break. Do so, it will help you to stick with the material, but don't forget to come back.

Subjects, that in my experience are particularly difficult to understand, I have repeated once or even more times, so the reader will be able to understand and remember them by the end of the book.

Some valuable introductory texts helped with writing this book; they are cited in the references, and recommended for further information, as a text of this size cannot cover everything. Indeed, it is not the objective of this book to represent the 'be all and end all' of Magnetic Resonance Imaging, but rather to serve as an appetizer for further reading.

Let us start with a general overview of MRI. .

The single steps of an MR examination can be described quite simply:

- the patient is placed in a magnet,
- a radio wave is sent in,
- the radio wave is turned off,
- the patient emits a signal, which is received and used for
- reconstruction of the picture

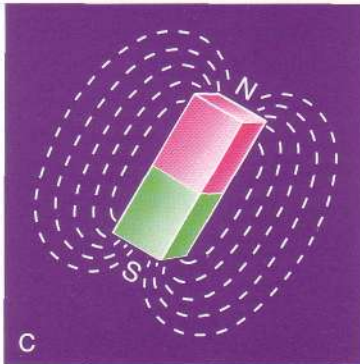
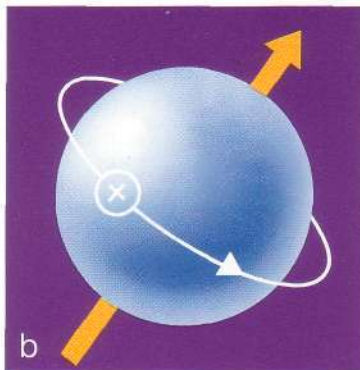
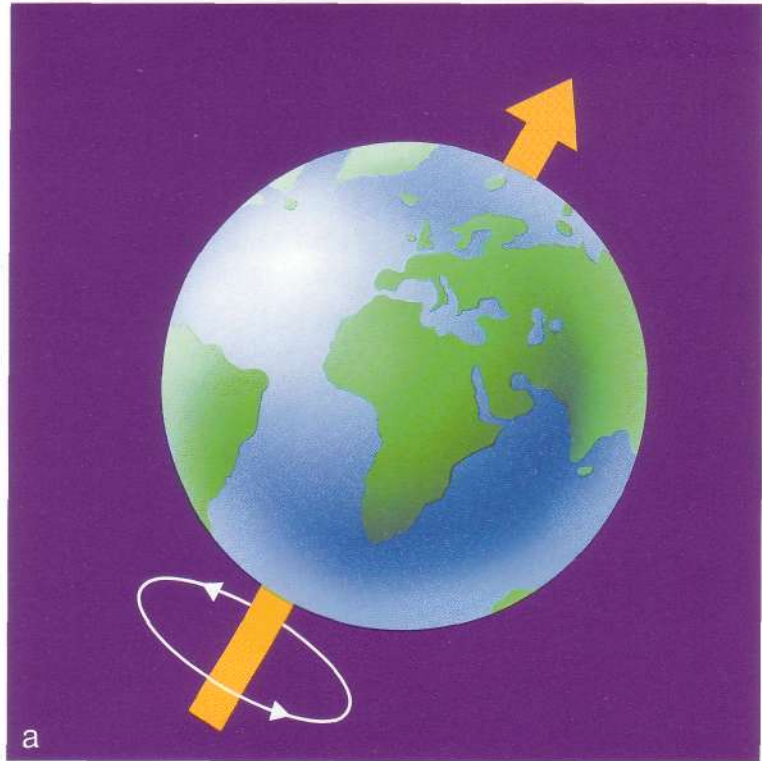
Let's take a look at these steps in detail

What happens, when we put a patient into the magnet of an MR machine?

To understand this, it is necessary to at least know some very basic physics - even though this may seem to be boring.

As we all know, atoms consist of a nucleus and a shell, which is made up of electrons. In the nucleus - besides other things - there are protons, little particles, that have a positive electrical charge (whatever that may actually be). These protons are analogous to little planets. Like earth, they are constantly turning, or spinning around an axis (fig. 1); or - as one says, protons possess a spin. The positive electrical charge, being attached to the proton, naturally spins around with it. And what is a moving electrical charge? It is an electrical current.

Now, may be you remember from your physics at school that an electrical current induces, causes a magnetic force, or magnetic field. So, where there is an electrical current, there is also a magnetic field.



This can be demonstrated very easily. Take a rusty nail and approach an electrical outlet - closer, closer. Do you feel it being repelled by the magnetic force so you do not put the nail into the outlet?

Fig. 1

Protons possess a positive charge. Like the earth they are constantly turning around an axis and have their own magnetic field.

Let's review what we have read

A proton has a spin, and thus the electrical charge of the proton also moves. A moving electrical charge is an electrical current, and this is accompanied by a magnetic field. Thus, the proton has its own magnetic field and it can be seen as a little bar magnet (fig. 1c).

What happens to the protons, when we put them into an external magnetic field?

The protons - being little magnets - align themselves in the external magnetic field like a compass needle in the magnetic field of the earth. However, there is an important difference. For the compass needle there is only one way to align itself with the magnetic field, for the protons however, there are two (fig. 2). The protons may align with their South and North poles

in the direction of the external field, parallel to it. Or they may point exactly in the complete opposite direction, anti-parallel. These types of alignment are on different energy levels.

To explain this; a man can align himself parallel to the magnetic field of the earth, i.e. walk on his feet, or he can align himself anti-parallel, in the opposite direction. Both states are on different energy levels, i.e. need different amounts of energy.

Walking on one's feet is undoubtedly less exhausting, takes less energy than walking on one's hands. (In the figures, this will be illustrated as pointing up or down, see fig. 2).

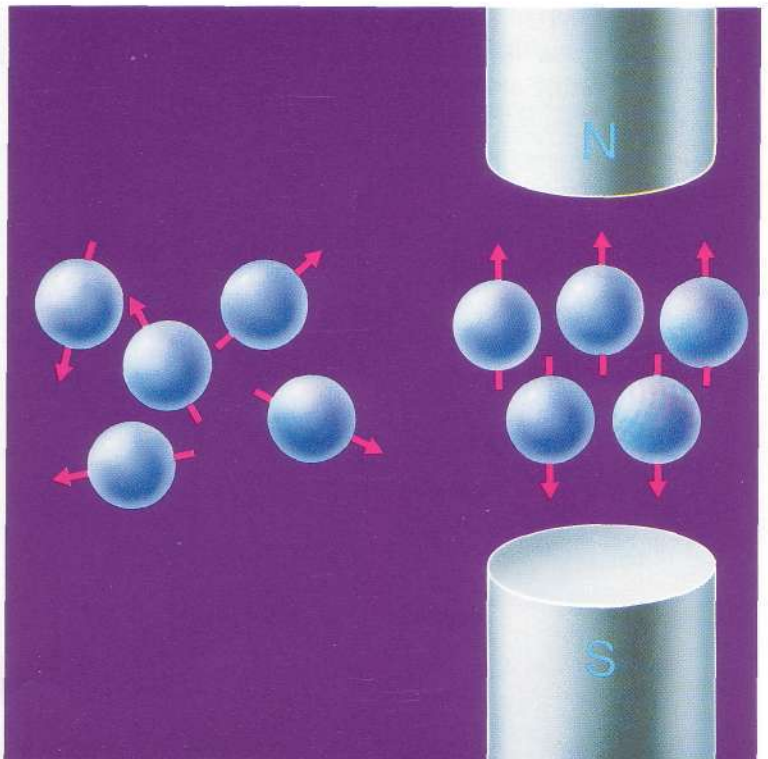


Fig. 2
Normally protons are aligned in a random fashion. This, however, changes when they are exposed to a strong external magnetic field. Then they are aligned in only two ways, either parallel or anti-parallel to the external magnetic field.

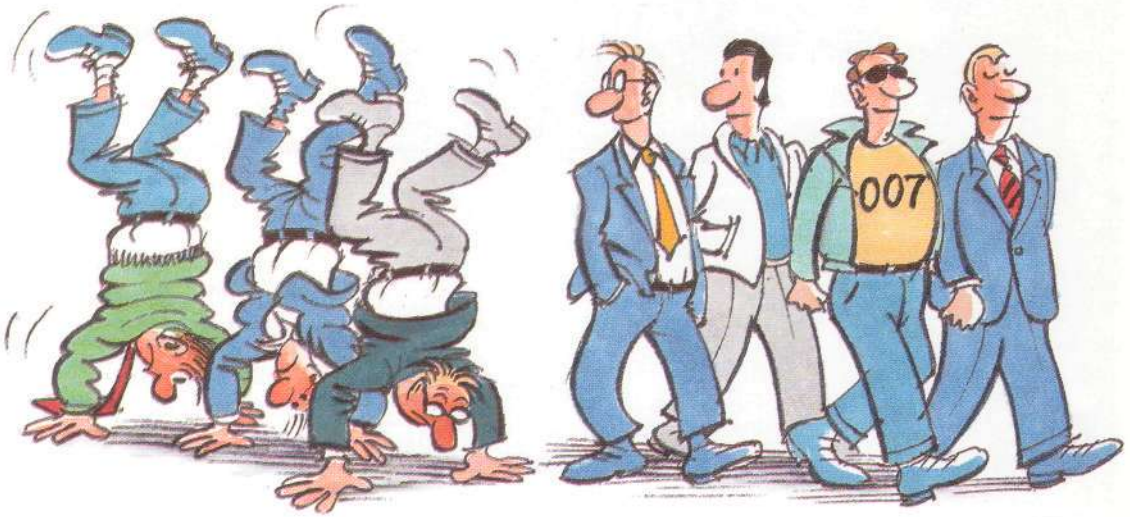


Fig. 3
When there are two possible states of alignment, the one that takes less energy, is on a lower energy level, is preferred.

Naturally the preferred state of alignment is the one that needs less energy. So more protons are on the lower energy level, parallel to the external magnetic field (walk on their feet). The difference in number is, however, very small and depends on the strength of the applied magnetic field. To get a rough idea: for about 10 million protons "walking on their hands", there are about 10 000 007 "walking on their feet" (the difference "007" is probably easy to remember).

It may be obvious at this point already, that for MRI the mobile protons are important (which are a subset of all protons that are in the body).

Let us take a closer look at these protons

We will see that the protons do not just lay there, aligned parallel or anti-parallel to the magnetic field lines. Instead, they move around in a certain way. The type of movement is called precession (fig. 4).

What type of movement is "precession"?

Just imagine a spinning top. When you hit it, it starts to "wobble" or tumble around. It does not, however, fall over. During this precession, the axis of the spinning top circles forming a cone shape (fig. 4).

It is hard to draw such a precessing proton, as this is a very fast movement as we will see below. For the sake of simplicity, we will just make "freeze frame" pictures, as if we were taking a fast flash light photograph of the situation at a specific moment in time.

For reasons we will learn below, it is important to know how fast the protons precess. This speed can be measured as precession frequency, that is, how many times the protons precess per second. This precession frequency is not constant. It depends upon the strength of the magnetic field (for magnetic field strength see page 96), in which the protons are placed. The stronger the magnetic field, the faster the precession rate and the higher the precession frequency.

This is like a violin string: the stronger the force exerted upon the string, the higher its frequency.

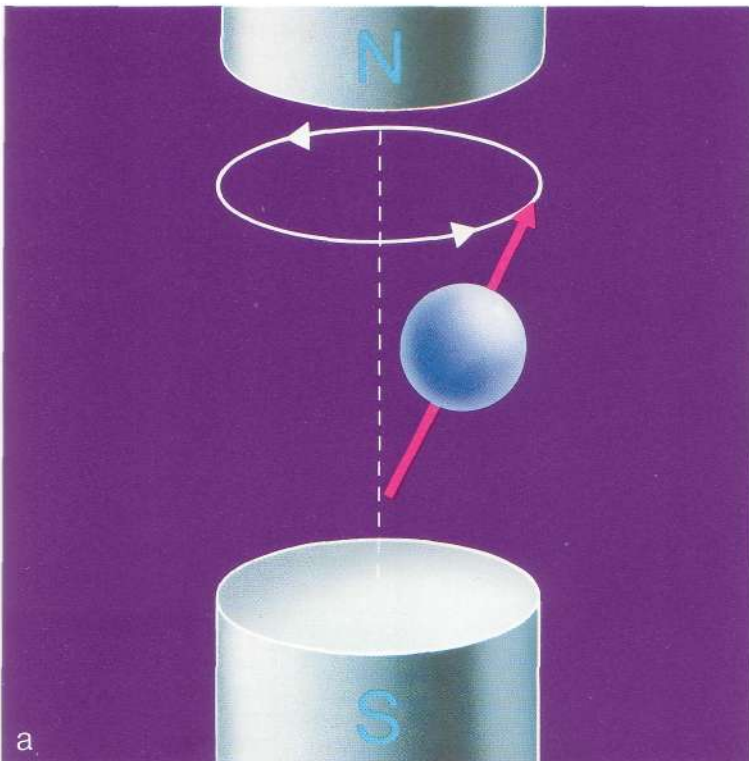
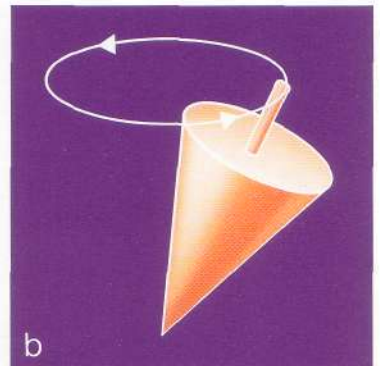


Fig. 4
A spinning top, which is hit, performs a wobbling type of motion. Protons in a strong magnetic field also show this type of motion, which is called precession.



It is possible and necessary to precisely calculate this frequency. This is done by using an equation called Larmor equation:

$$\omega_0 = \gamma B_0$$

ω_0 is the precession frequency (in Hz or MHz),

B_0 is the strength of the external magnetic field, which is given in Tesla (T) (see page 96), and

γ is the so-called gyro-magnetic ratio.

The equation states that the precession frequency becomes higher when the magnetic field strength increases. The exact relationship is determined by the gyro-magnetic ratio γ . This gyro-magnetic ratio is different for different materials (e.g. the value for protons is 42.5 MHz/T). It can be compared to an exchange rate, which is different for different currencies.

Time to take a break

however, let us briefly review, what we have read up to now:

- Protons have a positive electrical charge, which is constantly moving, because the protons possess a spin.
- This moving electrical charge is nothing more than an electrical current, and the latter always induces a magnetic field.
- So every proton has its little magnetic field, and can thus be seen as a little bar magnet.
- When we put a patient in the MR magnet, the protons, being little magnets, align with the external magnetic field. They do this in two ways: parallel and anti-parallel. The state that needs less energy is preferred, and so there are a few more protons "walking on their feet" than "on their hands" (fig. 3).
- The protons precess along the field lines of the magnetic field, just like a spinning top that precesses along the field lines of the magnetic field of earth.

• The precession frequency can be calculated by the Larmor equation, and is higher in stronger magnetic fields. Why is this precession frequency important?

It has something to do with the resonance in magnetic resonance imaging. But to understand this will take a few more minutes.

After the break you should go over the last summary again, and then continue . . .

Introducing the coordinate system

To make communication (and drawing of illustrations) easier, let us start using a coordinate system like we know from school (fig. 5). As you see, the z-axis runs in the direction of the magnetic field lines, and thus can represent them. So we can stop drawing the external magnet in all other illustrations.

From here on we will also illustrate the protons as vectors as little arrows. Maybe you remember: a vector represents a certain force (by its size) that acts in a certain direction (direction of the arrow). The force that is represented by vectors in our illustrations, is the magnetic force.

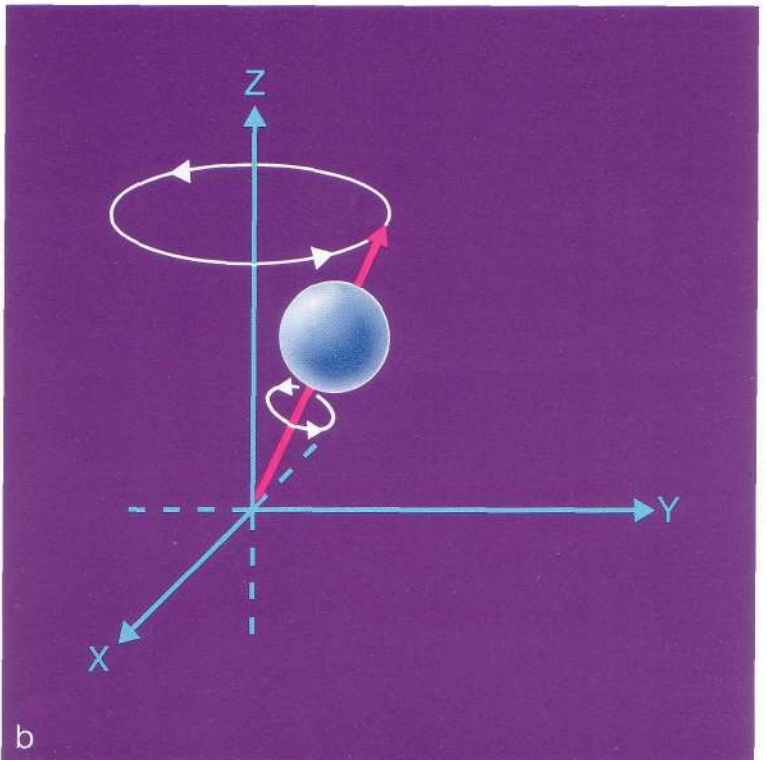
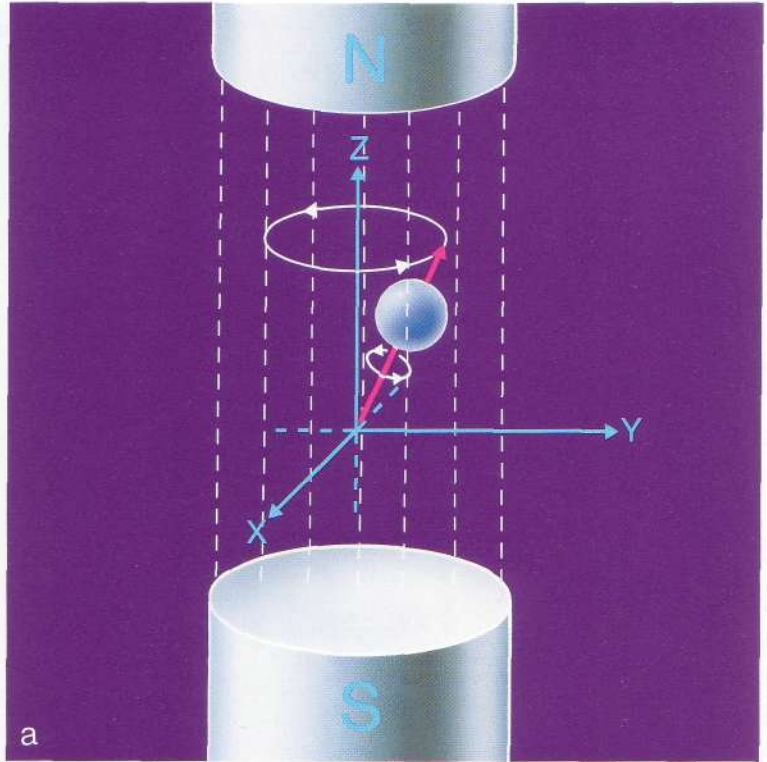


Fig. 5
Using a coordinate system makes the description of proton motion in the magnetic field easier, and also we can stop drawing the external magnet.

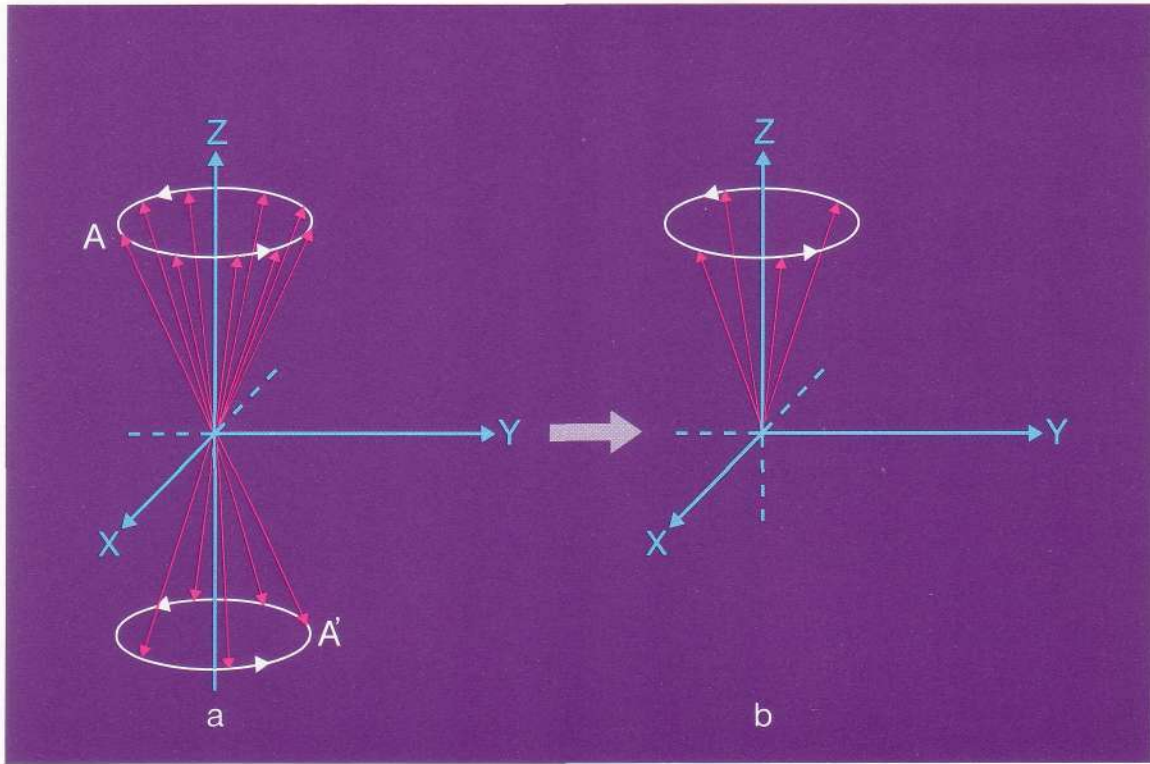


Fig. 6

The five protons, which "point" down cancel out the magnetic effects of the same number of protons, which "point" up (6a). So in effect it is sufficient to only look at the four unopposed protons (6b).

have read: there are more protons pointing up than down, and the magnetic forces of these protons are not cancelled by others. So we are left - in effect - with some protons (4 in our example) pointing up (fig. 6).

Now, look at figure 6. There we have 9 protons pointing up, precessing parallel to the external magnetic field lines, and 5 protons pointing down, precessing anti-parallel to the external magnetic field. As we stated above, what we see in the figure is just a picture taken at a specific point in time. A picture taken just a little later would show the protons in different positions because they precess. The precession actually goes very fast, the precession frequency for hydrogen protons is somewhere around 42 MHz in a magnetic field strength of 1 Tesla (see page 96); this means that the protons precess around the

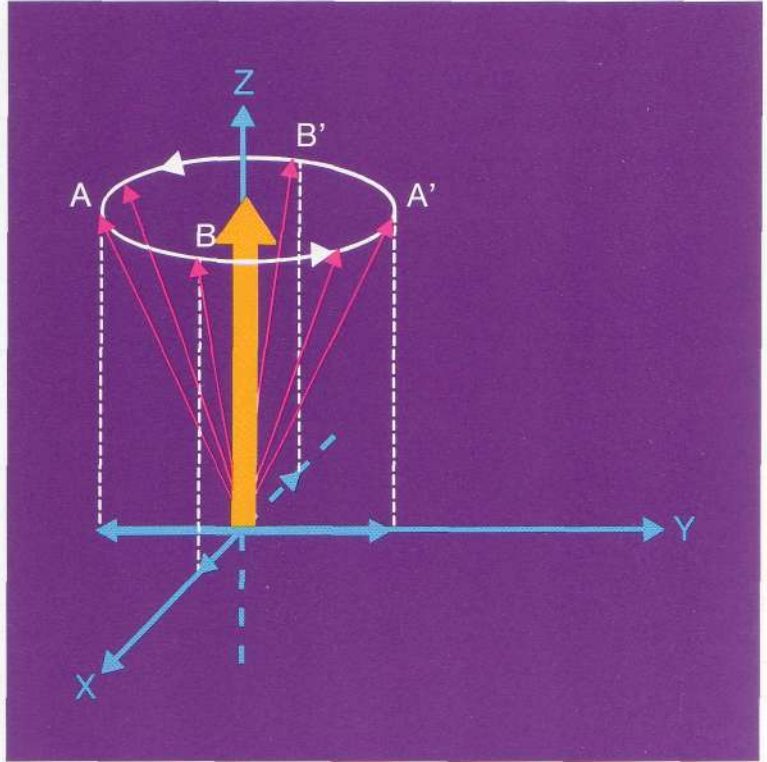
"ice cream cone" more than 42 million times per second. Now there are millions and millions of protons in your body precessing this fast. It is easy to imagine, that at a certain moment, there may be one proton (A in the illustration) pointing in one direction, and another proton (A') pointing exactly in the opposite direction. The result is very important; the magnetic forces in the opposing directions cancel each other out, like two persons pulling at the opposite ends of a rope. Finally, for every proton pointing down, there is one pointing up, cancelling its magnetic effects. But as we

However, not only magnetic forces pointing up and down can cancel or neutralize each other. As the protons that are left pointing up, precess, there may be one pointing to the right, when there is another one pointing to the left; or for one pointing to the front, there is one pointing backwards, and so on (the corresponding protons in fig. 7 are marked A and A', B and B' for example)

This means that the opposing magnetic forces of the remaining protons cancel each other out in these directions. This is true for all but one direction, the direction of the z-axis, along the external magnetic field (fig. 7). In this direction, the single vectors, the single magnetic forces add up, like people pulling on the same end of a rope.

What we end up with in effect is a magnetic vector in the direction of the external magnetic field (the arrow on the z-axis in fig. 7); and this vector is a sum vector made up by adding the magnetic vectors of the protons pointing upwards.

Now - what does this mean? This means that by placing a patient in the magnet of the MR unit (or in any other strong magnetic field), the patient himself becomes a magnet, i.e. has



his own magnetic field. Why? Because the vectors of the protons, that do not cancel each other out, add up (fig. 8).

As this magnetization is in direction along/longitudinal to the external magnetic field, it is also called longitudinal magnetization.

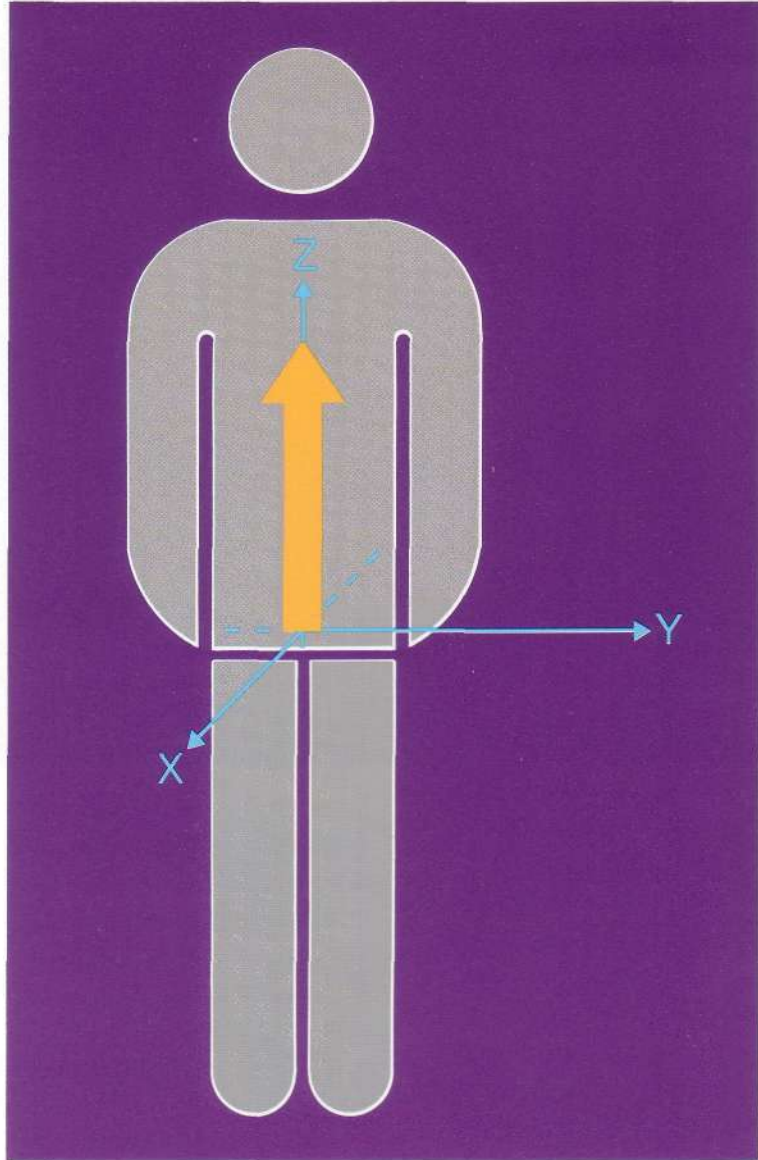
Fig. 7

The magnetic force of proton A, illustrated as an arrow, a vector, can be looked at as resulting from two components: one pointing up along the z-axis, and one in direction of the y-axis. The component along the y-axis is cancelled out by proton A', the magnetic force of which also has a component along the y-axis, however, in the opposite direction. The same holds true for other protons, e.g. B and B', which cancel their respective magnetic vectors along the x-axis. In contrast to the magnetic vectors in the x-y-plane, which cancel each other out, the vectors along the z-axis point in the same direction, and thus add up to a new magnetic sum vector pointing up.

Fig. 8

In a strong external magnetic field a new magnetic vector is induced in the patient, who becomes a magnet himself. This new magnetic vector is aligned with the external magnetic field.

As we have seen, the resulting new magnetic vector of the patient points in the direction of the external field, along its field lines. This is described as longitudinal direction. And it is actually this new magnetic vector that may be used to get a signal. It would be nice if we could measure this magnetization of the patient, but there is a problem: we cannot measure this magnetic force, as it is in the same direction, parallel to the external magnetic field (figs. 7 and 8).

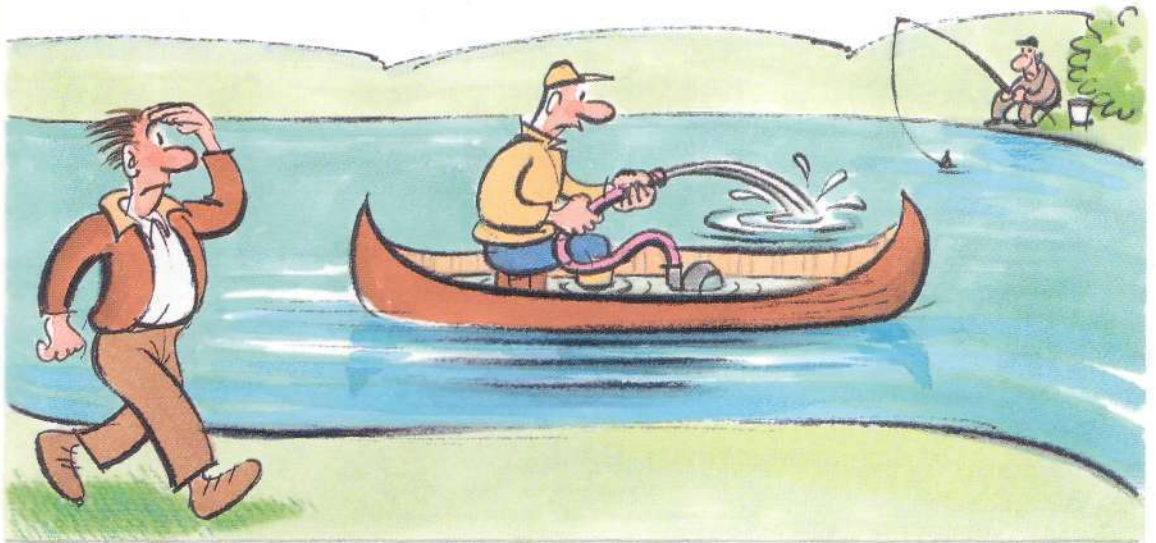


To illustrate this: imagine that you are standing on a boat, floating down a river. You have a water hose in your hand and squirt water into the river. For somebody who is watching you from the shore, it is impossible to tell how much water you pour out (this shall be how much new magnetization is added in the old direction).

However, when you point the water hose to the shore, change the direction of the new magnetic field, then the water may perhaps be directly picked up and measured by an impartial observer on the shore (fig. 9). What we should learn from this is: magnetization along or, better, longitudinal to the external magnetic field cannot be

measured directly. For this we need a magnetization which is not longitudinal, but transversal to the external magnetic field.

Fig. 9
Magnetization along an external magnetic field cannot be measured. For this a magnetization transverse to the external magnetic field is necessary.



Time to take a break . . .

but before you walk off, just read the short summary. And when you come back, start out with the summary again.

- Protons have a positive charge and possess a spin. Due to this, they have a magnetic field and can be seen as little bar magnets.
- When we put them into a strong external magnetic field, they align with it, some parallel (pointing up), some anti-parallel (pointing down).
- The protons do not just lay there, but precess around the magnetic field lines. And the stronger the magnetic field, the higher the precession frequency, a relationship that is mathematically described in the Larmor equation.

- Anti-parallel and parallel protons can cancel each others forces out. But as there are more parallel protons on the lower energy level ("pointing up"), we are left with some protons, the magnetic forces of which are not cancelled. All of these protons pointing up, add up their forces in the direction of the external magnetic field. And so when we put the patient in the MR magnet, he has his own magnetic field, which is longitudinal to the external field of the MR machine's magnet (figs. 7 and 8). Because it is longitudinal however, it cannot be measured directly.

What happens after we put the patient into the magnet?

We send in a radio wave. The term radio wave is used to describe an electromagnetic wave, that is in the frequency range of the waves which you receive in your radio. What we actually send into the patient is not a wave of long duration, but a short burst of some electromagnetic wave, which is called a radio frequency (RF) pulse. The purpose of this RF pulse is to disturb the protons, which are peacefully precessing in

alignment with the external magnetic field. Not every RF pulse disturbs the alignment of the protons. For this, we need a special RF pulse, one that can exchange energy with the protons.

This is as if someone were looking at you. You may not notice it, because there is no exchange of energy, so you do not change your position/alignment. However, if someone were to pound you in the stomach, exchange energy with you, your alignment would be disturbed. And this may explain why we need a certain RF pulse that can exchange energy with the protons to change their alignment.

Fig. 10 a + b
Energy exchange is possible when protons and the radiofrequency pulse have the same frequency.

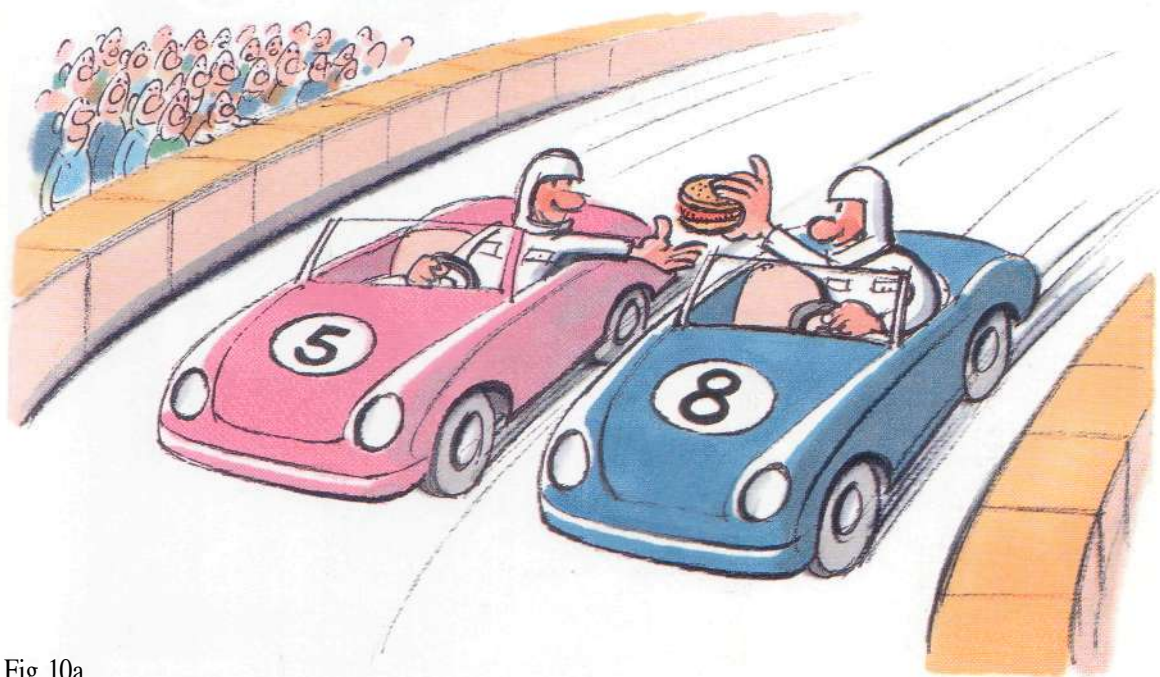


Fig. 10a

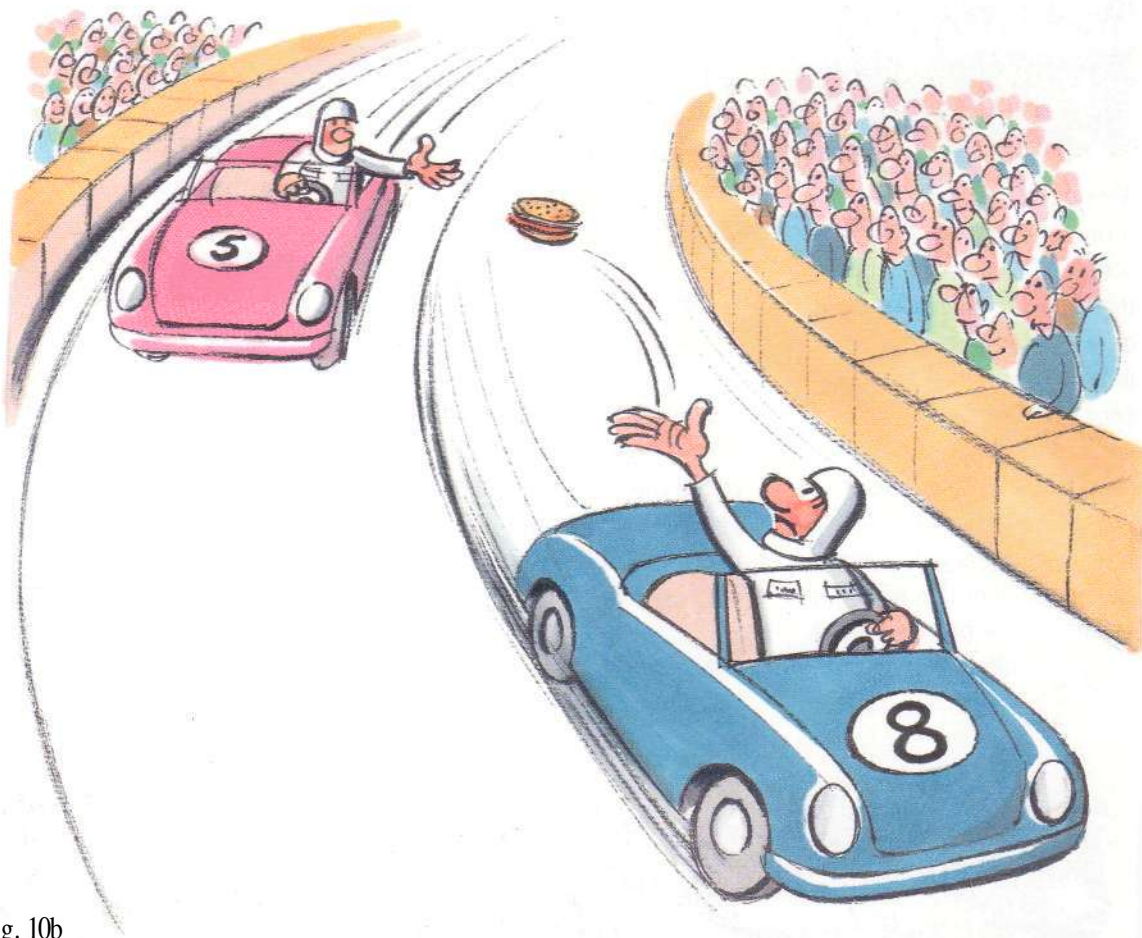


Fig. 10b

But when can an RF pulse exchange energy with the protons? For this it must have the same frequency; the same "speed" as the protons.

Just imagine that you are driving down a race track in your car, and someone in the lane next to you wants to hand you a couple of sandwiches, i.e. exchange energy with you (as you are hungry, the sandwiches would give you new energy). This energy trans-

fer is possible when both cars have the same speed, move around the race track with the same frequency.

With differences in speed/frequency less or no energy transfer is possible (fig. 10b).

Fig. 11

The radiofrequency pulse exchanges energy with the protons (a), and some of them are lifted to a higher level of energy, pointing downward in the illustration (b). In effect the magnetization along the z-axis decreases, as the protons which point down "neutralize" the same number of protons pointing up.

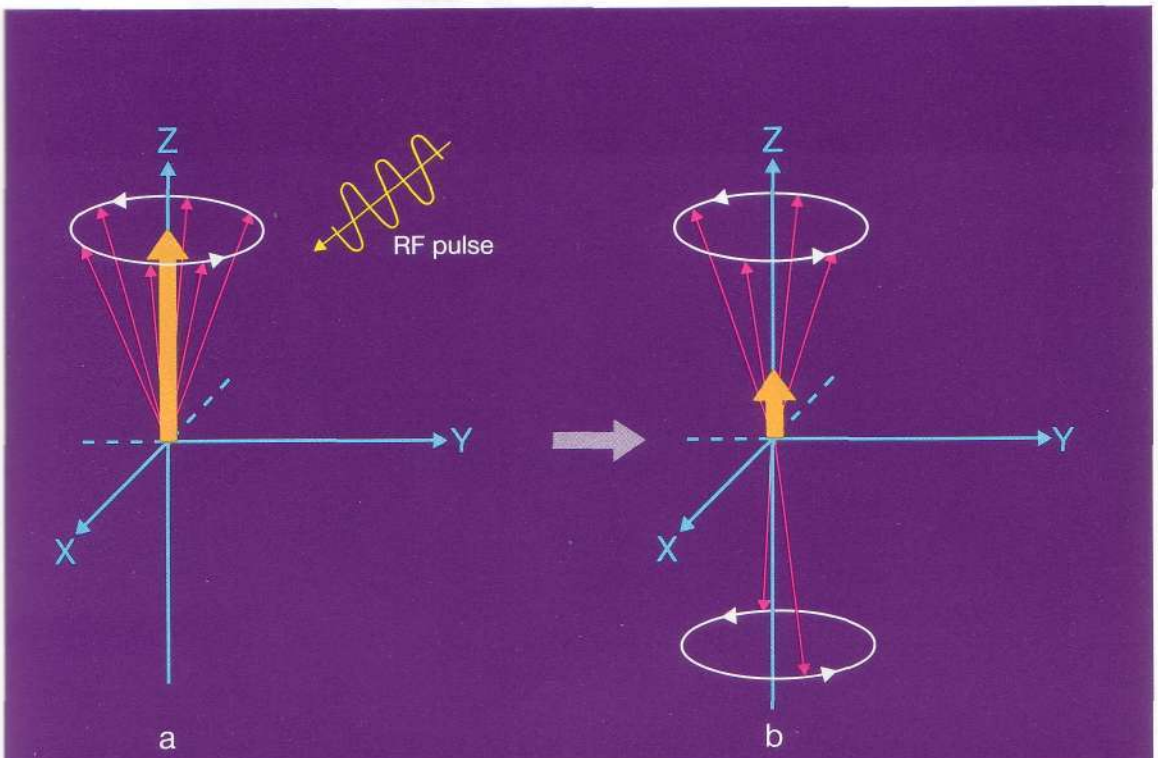
What speed, or better, what frequency did the protons have?

They had their precession frequency which can be calculated by the Larmor equation (see page 10). So the Larmor equation gives us the necessary frequency of the RF pulse to send in. Only when the RF pulse and the protons have the same frequency, can protons pick up some energy from the radio wave, a phenomenon called resonance (this is where the "resonance" in magnetic resonance comes from).

The term resonance can be illustrated by the use of tuning forks. Imagine that you are in a room with different kinds of tuning forks, tuned e.g. to a, e, and d. Somebody enters the room with a tuning fork with "a"-frequency, that was struck to emit sound. From all the tuning forks in the room, all of a sudden the other "a" forks, and only those, pick up energy, start to vibrate and to emit sound, they show a phenomenon called resonance.

What happens with the protons, when they are exposed to this RF-pulse?

Some of them pick up energy, and go from a lower to a higher energy level. Some, which were walking on their feet, start walking on their hands. And this has some effect on the patients magnetization, as you can see in figure 11. Let us assume that from the net sum of 6 protons pointing up, after the RF pulse is sent in 2 point down. The result is that these 2 protons cancel out the magnetic forces of the same number of protons,



that point up. In effect, then, the magnetization in longitudinal direction decreases from 6 to 2.

But something else happens. Do you remember what drawings of radio waves look like? Just look at fig. 12; they resemble a whip, and the RF pulse also has a whip-like action (fig. 13): it gets the precessing protons in synch and this has another important effect.

When the protons randomly point left/right, back/forth and so on, they also cancel their magnetic forces in these directions (as we read on page 13). Due to the RF pulse, the protons do not point in random directions any more, but move in step, in synch - they are "in phase". They now point in the same direction at the same time, and thus their magnetic vectors add up in this direction.

This results in a magnetic vector pointing to the side to which the precessing protons point, and this is in a transverse direction (—• fig. 13). This is why it is called transversal magnetization.



Fig. 12
The drawing of radiowaves normally resembles a whip, and radiowaves in MRI also have a whip-like action.

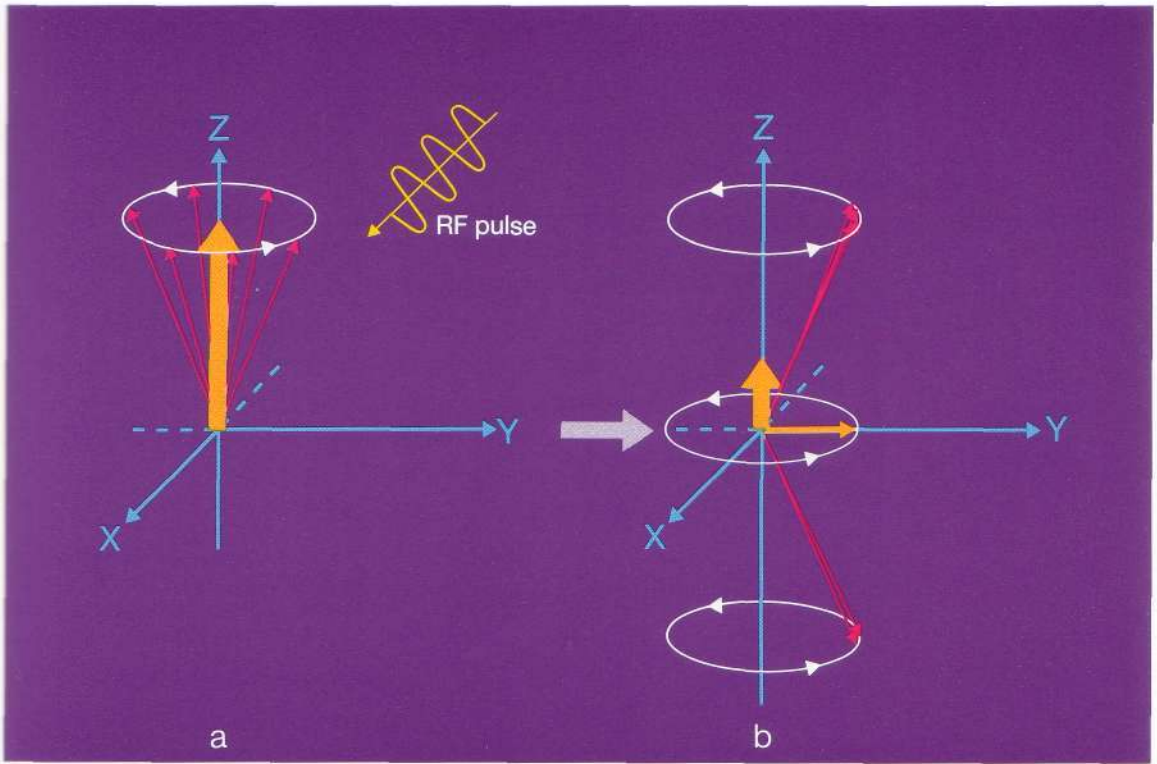
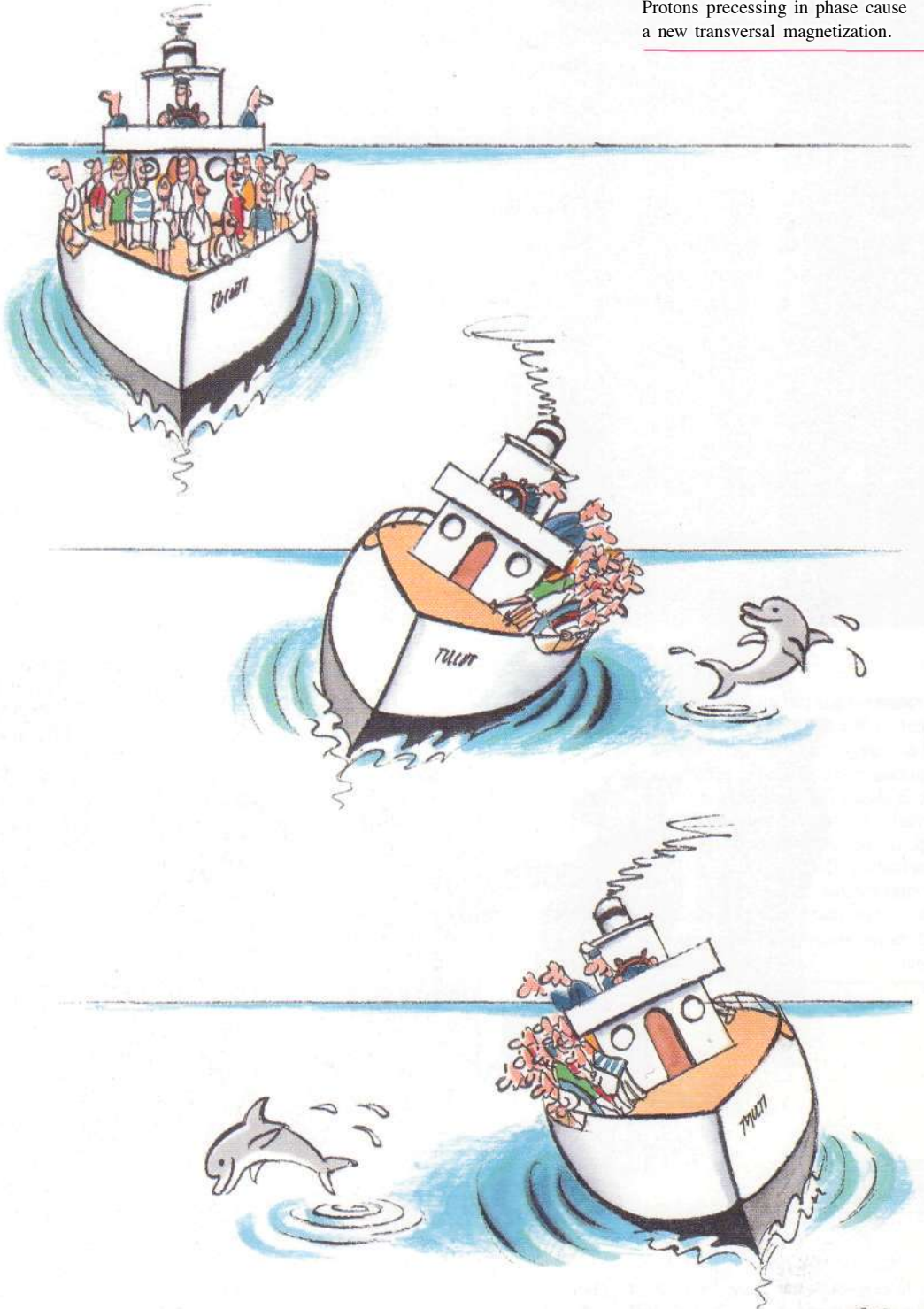


Fig. 13
 The radiowave has two effects on the protons: it lifts some protons to a higher level of energy (they point down), and it also causes the protons to precess in step, in phase. The former results in decreasing the magnetization along the z-axis, the so-called longitudinal magnetization. The latter establishes a new magnetization in the x-y-plane (—•), a new transversal magnetization, which moves around with the precessing protons.

The situation can be compared to a ship: think about the passengers being distributed randomly all over the deck, the ship then is in a normal position. Then have all passengers walk in equal step around the railing; what happens? The ship is leaning towards the side where the people are, a new force is established, and becomes visible (fig. 14).

So the RF pulse causes a transversal magnetization. This newly established magnetic vector naturally does not stand still, but moves in line with the precessing protons, and thus with the precession frequency, (fig. 13).

Fig. 14
Protons precessing in phase cause
a new transversal magnetization.



So - what were the new things that we have learned?

Repeat them using fig. 15!

• When we put the patient in the MR machine, the protons line up parallel or anti-parallel to the machine's magnetic field. A magnetic field in the patient, longitudinal to the external field results (fig. 15a).

fig. 15

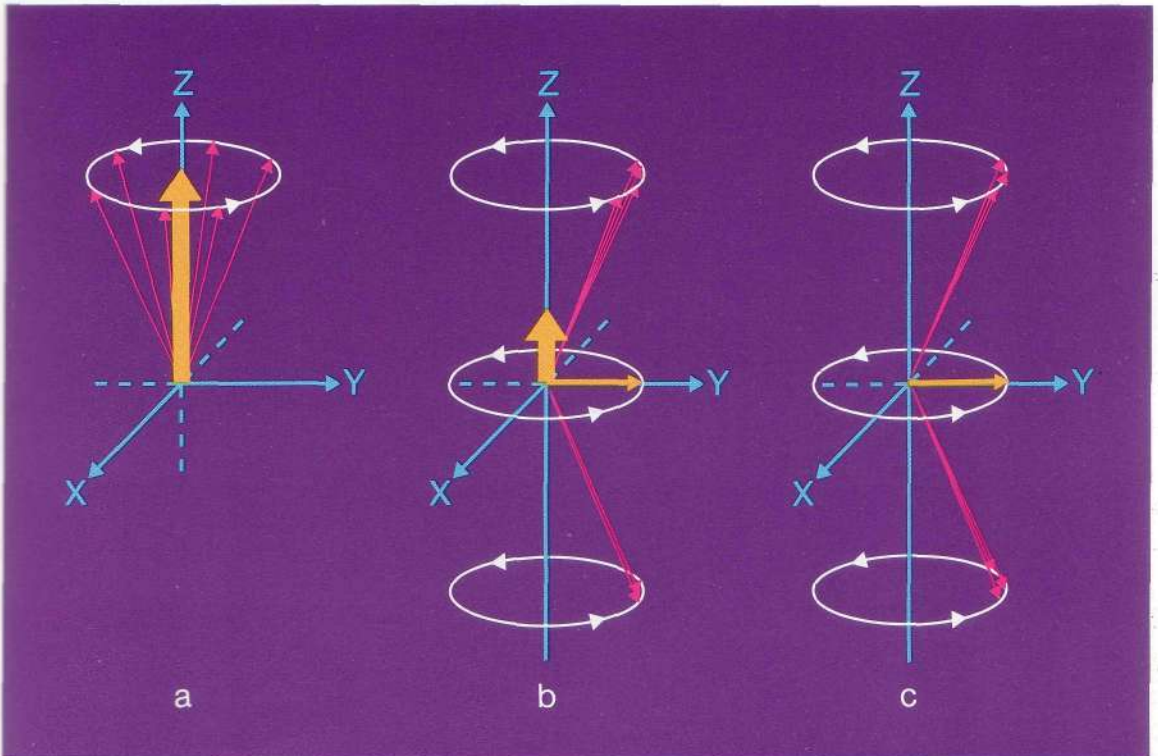
In a strong external magnetic field a new magnetic vector along the external field is established in the patient (a). Sending in an RF pulse causes a new transversal magnetization while longitudinal magnetization decreases (b). Depending on the RF pulse, longitudinal magnetization may even totally disappear (c).

• Sending in an RF pulse that has the same frequency as the precession frequency of the protons causes two effects:

- Some protons pick up energy, start to walk on their hands, and thus decrease the amount of longitudinal magnetization.

- The protons get in synch, start to precess in phase. Their vectors now also add up in a direction transverse to the external magnetic field, and thus a transversal magnetization is established.

In summary: the RF pulse causes longitudinal magnetization to decrease, and establishes a new transversal magnetization (figs. 13 and 15).



Let us have a look at that newly established transversal magnetization vector

This moves in phase with the precessing protons (fig. 16). When you watch what is happening from outside, the new magnetic vector comes towards you, goes away from you, comes again towards you, and so on. And this is important: the magnetic vector, by constantly moving, constantly changing, induces an electric current. We talked about the opposite already: the moving electrical

charge of the proton, the electrical current, induces the proton's magnetic field.

This also is true the other way around: a moving magnetic field causes an electrical current, e.g. in an antenna, as is the case with TV/radio waves.

(The term electromagnetic field should actually remind us of the relationship between electrical current and magnetism). As we learned above we also have a moving/changing magnetic vector in MRI. This can also induce an electrical current in an antenna, which is the MRI signal.

As the transversal magnetic vector moves around with the precessing protons, it comes to-

wards the antenna, goes away from it, comes towards it again and so on, also with the precession frequency. The resulting MR signal therefore also has the precession frequency (fig. 16).

But . . . how can we make a picture out of this electrical current, which actually is our MR signal?

For this we have to know where in the body the signal came from. How can we know that? The trick is really quite simple: we do not put the patient into a magnetic field which has the same strength all over the section of the patient, which we want to examine. Instead we take a magnetic field, which has a different strength at each point of the patients cross section. What does this do?

We heard that the precession frequency of a proton depends on the strength of the magnetic field (as the frequency of a violin string depends on the strength with which you pull it).

If this strength is different from point to point in the patient, then protons in different places



Fig. 16

The new transversal magnetization moves around with the precessing protons (see fig. 7). Thus for an external observer, transversal magnetization constantly changes its direction, and can induce a signal in an antenna.

process with different frequencies. And as they precess with different frequencies, the resulting MR signal from different locations also has a different frequency. And by the frequency we can assign a signal to a certain location.

It is like with your TV: when you are in the kitchen (where you probably do not have a TV) and hear a sound from your favourite TV show, you know where the sound is coming from. It comes from that spot in your apartment where the TV stands. What you subconsciously do, is connect a certain sound to a certain location in space.

This shall suffice for spatial information right now, on page 89 we will go into more detail about this.

Further details about the MR signal

If our protons rotated around in synch, in phase, and nothing would change, then we would get a signal as it is illustrated in fig. 16. This, however, is not what happens. As soon as the RF pulse is switched off, the whole system, which was disturbed by the RF pulse, goes back to its original quiet, peaceful state, it relaxes. The newly

established transverse magnetization starts to disappear (a process called transversal relaxation), and the longitudinal magnetization grows back to its original size (a process called longitudinal relaxation). Why is that?

The reason why the longitudinal magnetization grows back to its normal size is easier to explain, so let us start with that (see fig. 17).

No proton walks on its hands longer than it has to - sort of a human trait. The protons that were lifted to a higher energy level by the RF pulse go back to their lower energy level, and start to walk on their feet again.



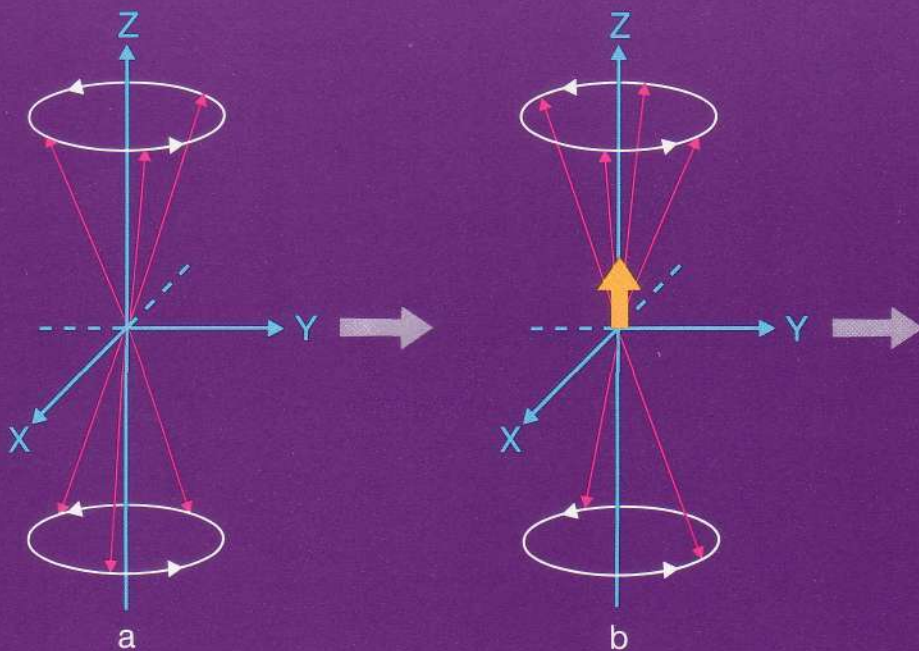


Fig. 17

After the RF pulse is switched off, protons go back from their higher to the lower state of energy, i.e. point up again. This is illustrated "one-by-one".

The effect is that longitudinal magnetization increases and grows back to its original value. Note that for simplicity the protons were not depicted as being in phase; this subject is covered in more detail in fig. 20 and 26.

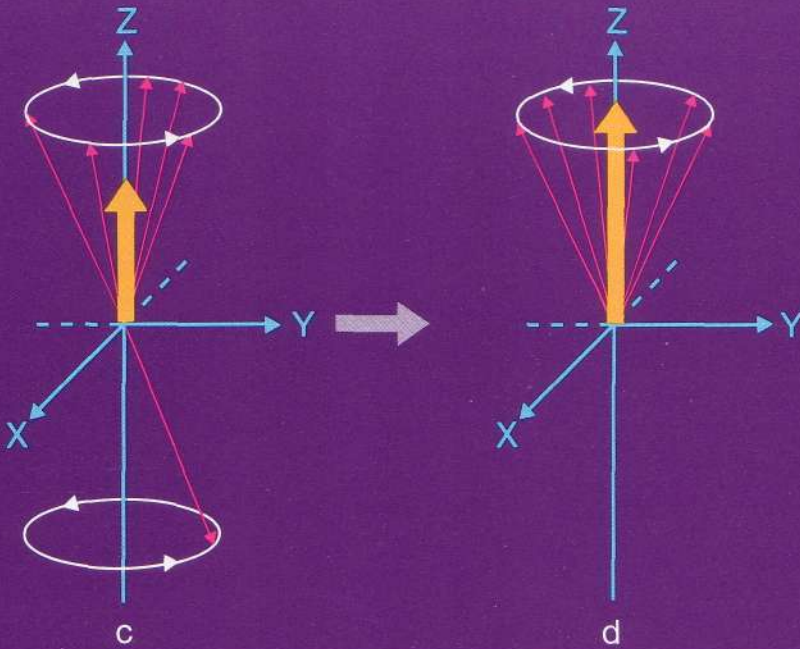
Not all protons do this at the very same time, instead it is a continuous process, as if one proton after the other goes back to its original state.

This is illustrated in fig. 17 for a group of protons. For the sake of simplicity the protons are shown as being out of phase, which of course they aren't in the beginning.

Why and how they stop precessing in phase will be explained a little later.

What happens to the energy which they had picked up from the RF pulse?

This energy is just handed over to their surroundings, the so-called lattice. And this is why this process is not only called longitudinal relaxation, but also spin-lattice-relaxation.



By going back on their feet, pointing upwards again, these protons no longer cancel out the magnetic vectors of the same number of protons pointing up, as they did before. So, the magnetization in this direction, the longitudinal magnetization increases, and finally goes back to its original value (fig. 17).

If you plot the time vs. longitudinal magnetization, you get a curve like fig. 18, it increases with time. This curve is also called a T_1 -curve.

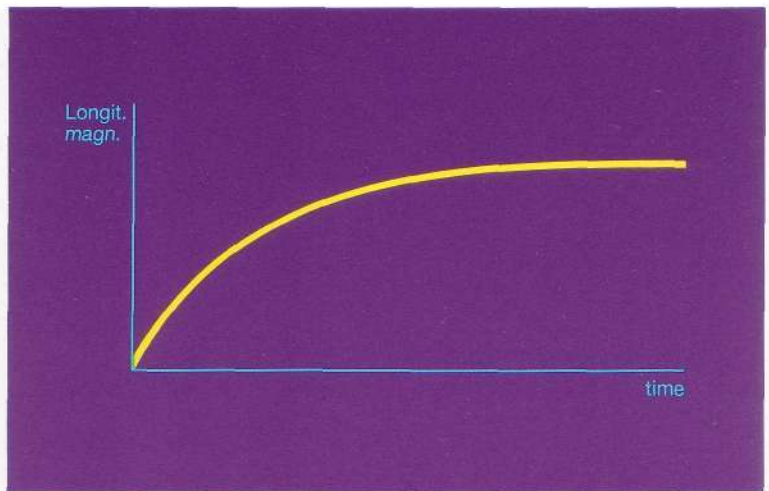


Fig. 18
If one plots the longitudinal magnetization vs. time after the RF pulse was switched off, one gets a so-called T_1 -curve.

Enough of the longitudinal magnetization - what happens with the transversal magnetization?

After the RF pulse is switched off, the protons get out of step, out of phase again, as nobody is telling them to stay in step. For the sake of simplicity this has been illustrated for a group of protons which all "point up" in fig.20.

We heard earlier that protons precess with a frequency which is determined by the magnetic

The time that it takes for the longitudinal magnetization to recover, to go back to its original value, is described by the longitudinal relaxation time, also called T_1 . This actually is not the exact time it takes, but a time constant, describing how fast this process goes.

This is like taking time for one round at a car race. The time gives you an idea of how long the race may take, but not the exact time. Or more scientifically, T_1 is a time constant comparable to the time constants that for example describe radioactive decay.

That T_1 is the longitudinal relaxation time can easily be remembered by looking at your typewriter: (fig. 19)

The "1" looks very much like the T_1 , reminding you also that it describes the spin- $\frac{1}{2}$ lattice relaxation. But there are more hidden hints to this: the "1" also looks like a match. And this match should remind you of something, which we also have mentioned already: longitudinal relaxation has something to do with exchange of energy, thermal energy, which the protons emit to the surrounding lattice while returning to their lower state of energy.

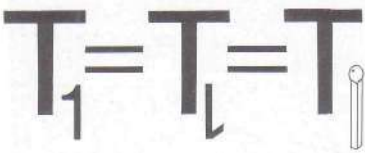
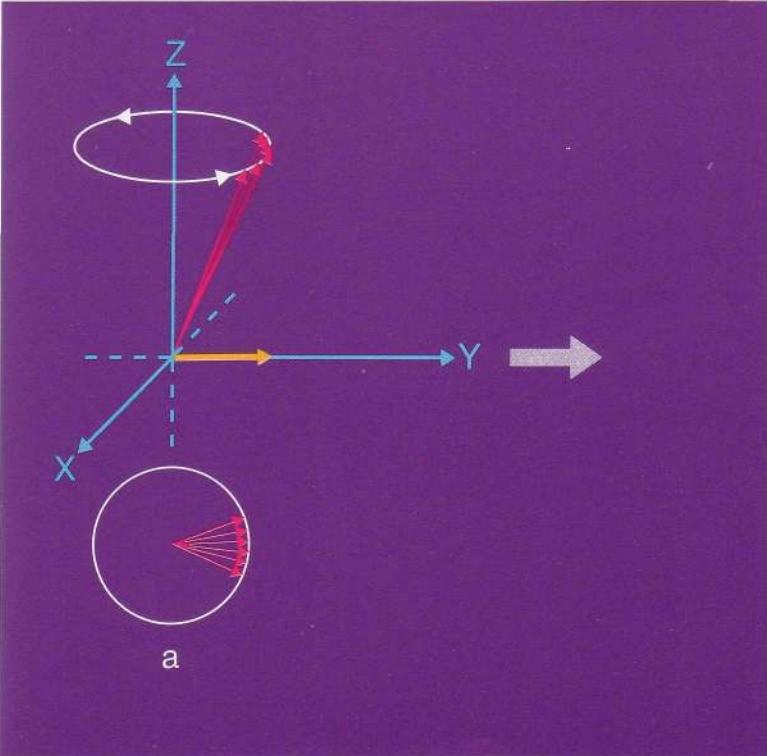


Fig. 19
 T_1 is the longitudinal relaxation time that has something to do with the exchange of thermal energy.



field strength that they are in. And all the protons should experience the same magnetic field. This, however, is not the case:

- the field of the MR magnet, in which the patient is placed, is not totally uniform, not totally homogenous, but varies a little, thus causing different precession frequencies
- and each proton is influenced by the small magnetic fields from neighbouring nuclei, that are also not distributed evenly, thus causing different precession frequencies too. These internal magnetic field variations are somehow characteristic for a tissue.

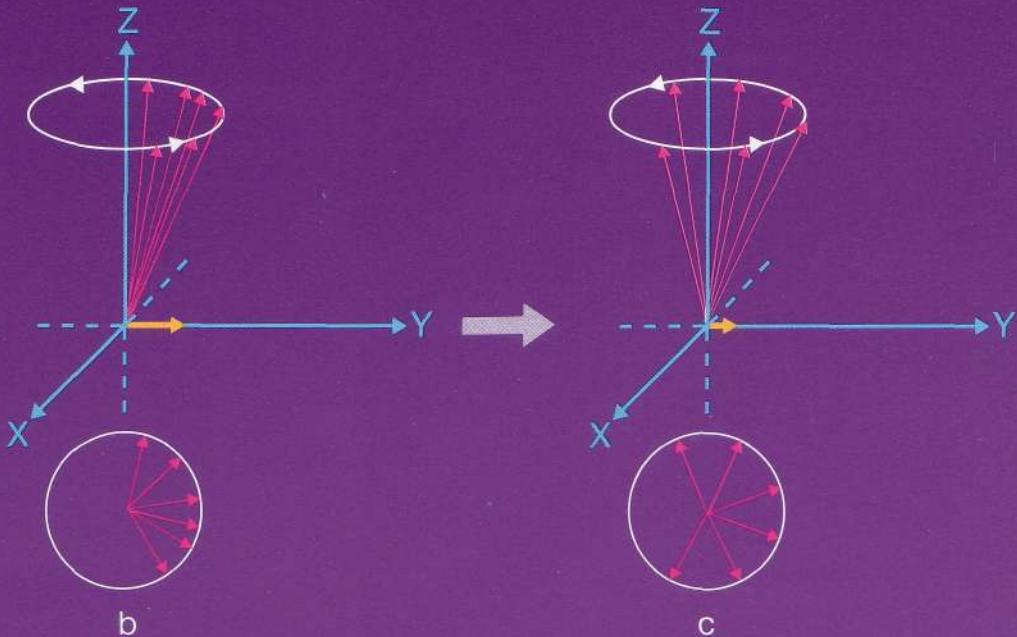
So after the RF pulse is switched off, the protons are no longer forced to stay in step; and as they have different precession frequencies, they will be soon out of phase.

It is interesting to see, how fast the protons get out of phase: just suppose that one proton (p_1) is rotating/precessing with a frequency of 10 megahertz, i.e. 10 million revolutions per second. Due to inhomogeneities a neighbouring proton (p_2) is in a magnetic field, which is 1 % stronger; this proton has a precession frequency of 10.1 megahertz, 1 % more. In 5 microseconds (0.000005 sec or 5×10^{-6}), p_1

will have made 50.5 turns/revolutions, while proton p_2 will have made only 50. So in this short time span, the protons will be 180° out of phase, cancelling their magnetic moments in the respective plane.

Fig. 20

After the RF pulse is switched off, protons lose phase coherence, they get out of step. When you look at these dephasing proton ensembles from the top (which is illustrated in the lower part of the figure), it becomes obvious, how they fan out. Fanning out, they point less and less in the same direction, and thus transversal magnetization decreases.



Similar to what we did for the longitudinal magnetization, we can plot transversal magnetization versus time. What we get is a curve like in figure 21. This curve is going downhill, as transversal magnetization disappears with time. And as you probably expect: there is also a time constant, describing how fast transversal magnetization vanishes, goes downhill. This time constant is the transversal relaxation time T_2 . How to remember what " T_2 " is?

Easy:

$$T_2 = T \times 2 = T T = T t,$$

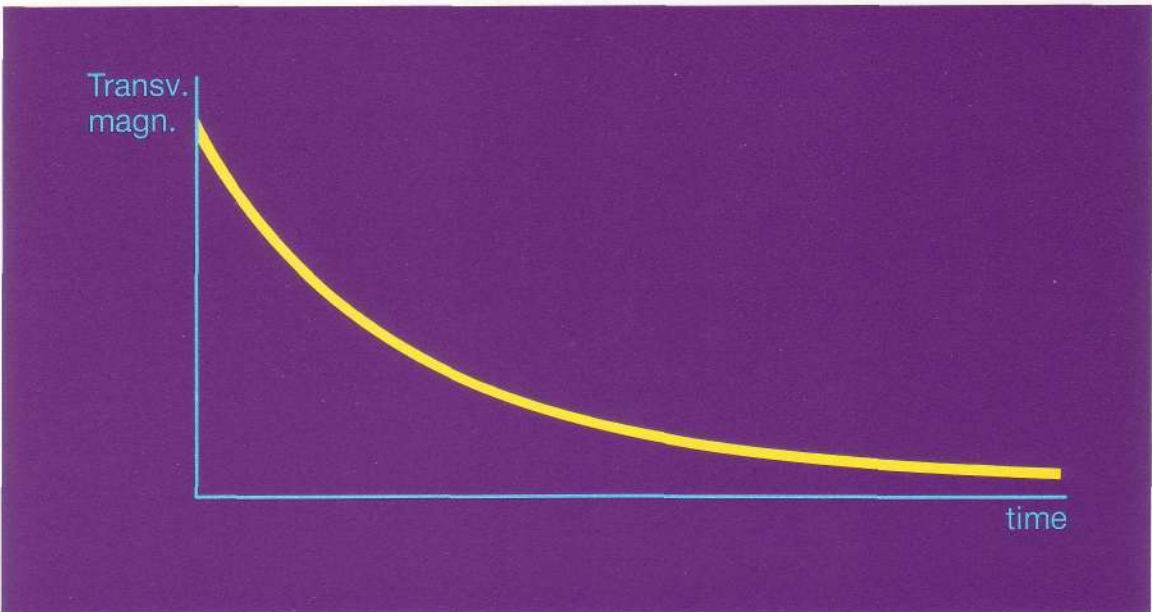
and this means, it describes the "T transversal", thus the relaxation of the transversal magnetization. The resulting curve in figure 21 thus is called a T_2 -curve. Another term for transversal relaxation is spin-spin-relaxation, reminding

us of the underlying mechanism, a spin-spin interaction.

How to remember, which one is the T_1 - and which the T_2 -curve? Just put both curves together, and you can see something like a mountain with a ski slope. You first have to go uphill (T_1 -curve), before you jump down (T_2 -curve) (fig. 22).

Fig. 21

If one plots transversal magnetization vs. time after the RF pulse is switched off, one gets a curve as illustrated, which is called a T_2 -curve.



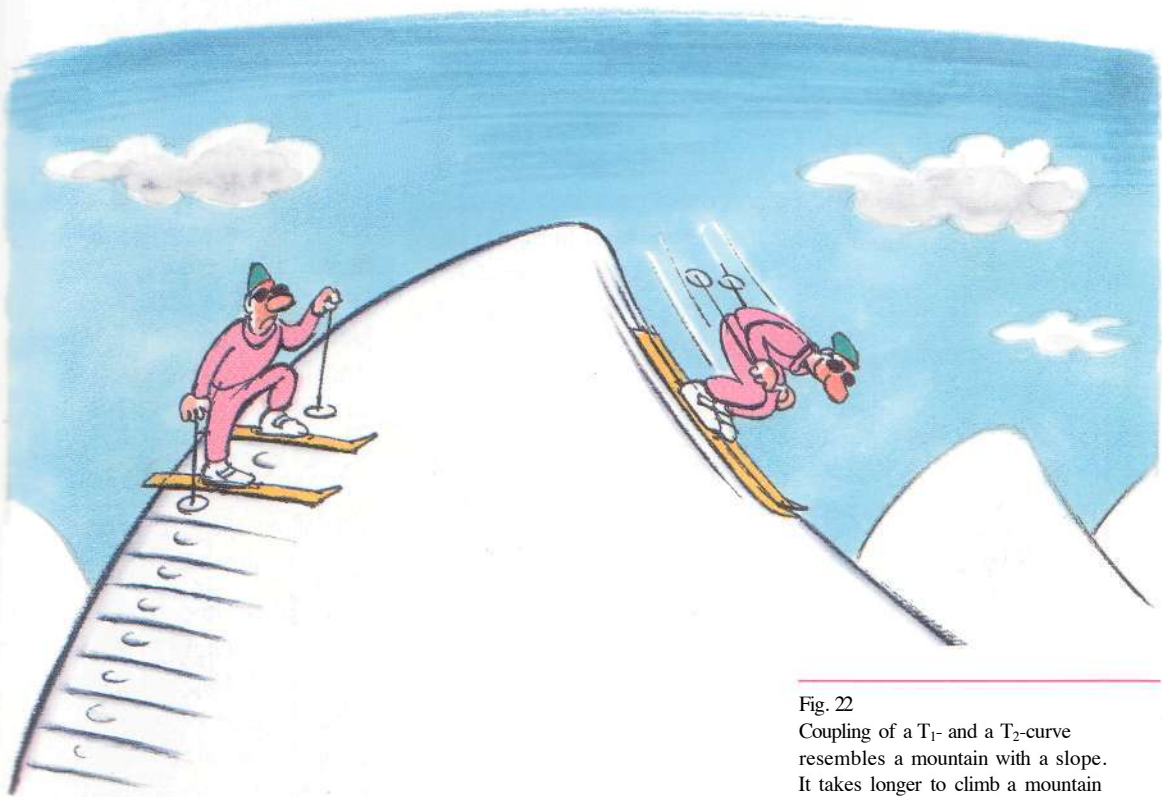
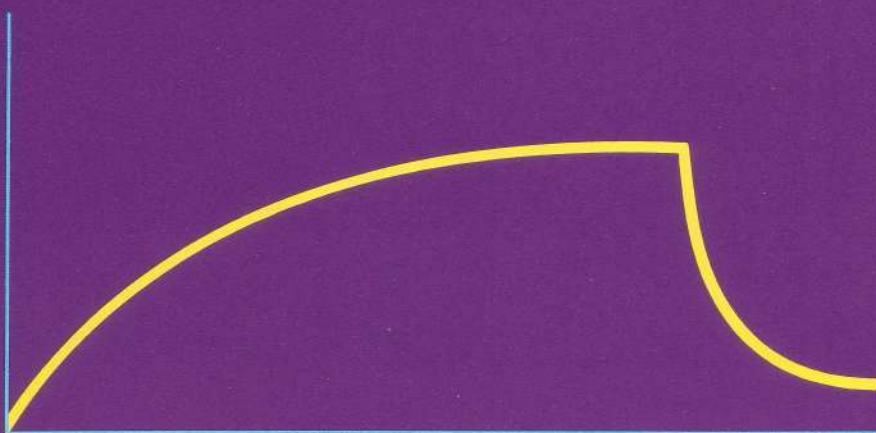


Fig. 22
Coupling of a T_1 - and a T_2 -curve
resembles a mountain with a slope.
It takes longer to climb a mountain
than to slide or jump down, which
helps to remember that T_1 is normally
longer than T_2 .

magn.



time

So, time to review

We have learned that

- protons are like little magnets
- in an external magnetic field they align parallel or anti-parallel
- the lower energy state (parallel) is preferred, so a few more protons align this way
- the protons perform a motion that resembles the wobbling of a spinning top, that was hit
- this motion is called precession
- the precession frequency is dependent on the strength of the external magnetic field (a relationship which is described by the Larmor equation).
The stronger the magnetic field, the higher the precession frequency.
- protons "pointing" in opposite directions cancel each other's magnetic effects in the respective directions
- as there are more protons aligned parallel to the external field, there is a net magnetic moment aligned with or longitudinal to the external magnetic field

- a radio frequency pulse that has the same frequency as the precessing protons, can cause resonance, transfer energy to the protons. This results in more protons being anti-parallel and thus neutralizing/cancelling more protons in the opposite direction. Consequence: the longitudinal magnetization decreases

- the RF pulse also causes the protons to precess in synch, in phase. This results in a new magnetic vector, the transversal magnetization

- when the RF pulse is switched off

- longitudinal magnetization increases again; this longitudinal relaxation is described by a time constant T_1 , the longitudinal relaxation time

- transversal magnetization decreases and disappears; this transversal relaxation is described by a time constant T_2 , the transversal relaxation time. Longitudinal and transversal relaxation are different, independent processes, and that is why we discussed them individually (see figs. 17 and 20).

This is what you should know by now.

How long is a relaxation time?

Look at our example with the T_1 - and T_2 -curves in fig. 22. It is probably easy and logical, that it takes you more time to get to the top of the mountain, than to go back down, to jump off. This means T_1 is longer than T_2 , and just to give you an idea: T_1 is about 2-5-10 times as long as T_2 . Or in absolute terms in biological tissues:

T_1 is about 300 to 2000 msec, and T_2 is about 30 to 150 msec.

It is difficult to pinpoint the end of the longitudinal and transversal relaxation exactly. Thus, T_1 and T_2 were not defined as the times when relaxation is completed. Instead T_1 was defined as the time when about 63% of the original longitudinal magnetization is reached. T_2 is the time when transversal magnetization decreased to 37% of the original value. These percentages are derived from mathematical equations ($63\% = 1 - 1/e$; $37\% = 1/e$) describing signal intensity, but we do not want to go into more detail here.

(However, we should mention that $1/T_1$ is also called longitudinal relaxation rate, and $1/T_2$ transversal relaxation rate).

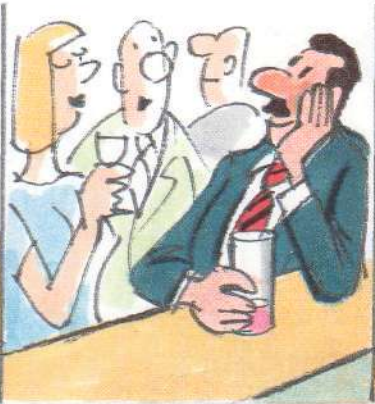
Previously it was believed that measuring the relaxation times, would give tissue characteristic results, and thus enable exact tissue typing. This, however, proved to be wrong, as there is quite some overlap of time ranges; and also T_1 is dependent on the magnetic field strength used for the examination (see page 36).

What is a long/short relaxation time, and which tissues have long/short relaxation times? Look at fig. 23 - what do you see? You see somebody having a long drink, something liquid (representing water). When you go to your favourite bar (which naturally is crowded, as it is a popular place) and order a long drink, you have to

wait quite a while to get your drink - T_1 is long. When you finally have your long drink, it takes you also a long time to drink it, so T_2 also is long. And we want to remember: water/liquids have a long T_1 and a long T_2 .

Fig. 23
Liquids have a long T_1 and a long T_2 .





Now look at fig. 24, you see somebody getting a hamburger. These normally contain much fat, and shall represent fat for us. The hamburger is fast food, you get it fast, thus fat has a short T_1 . What about T_2 ? It takes some time to eat fast food, fat; however, you normally spend more time with your long drink, so fat has a shorter T_2 than water.

As water has a long T_1 and a long T_2 , it is easy to imagine that "watery tissues", tissues with a high water content, can also have long relaxation times. Interestingly enough, pathological/diseased tissues often have a higher water content than the surrounding normal tissues.

Fig. 24
Compared to liquids/water fat has a short T_1 and a short T_2 .



What is T_1 influenced by?

Actually, T_1 depends on tissue composition, structure and surroundings. As we have read, T_1 -relaxation has something to do with the exchange of thermal energy, which is handed over from the protons to the surroundings, the lattice. The precessing protons have a magnetic field, that constantly changes directions, that constantly fluctuates with the Larmor frequency. The lattice also has its own magnetic fields. The protons now want to hand energy over to the lattice to relax. This can be done very effectively, when the fluctuations of the magnetic fields in the lattice occur with a frequency that is near the Larmor frequency.

When the lattice consists of pure liquid/water, it is difficult for the protons to get rid of their energy, as the small water molecules move too rapidly. And as the protons (which are on the higher energy level) cannot hand their energy over to the lattice quickly, they will only slowly go back to their lower energy level, their longitudinal alignment. Thus it takes a long time for the longitudinal magnetization to show up again, and this means that liquids/water have long T_1 s.

When the lattice consists of medium-size molecules (most body tissues can be looked at as liquids containing various sized molecules, kind of like a soup), that move and have fluctuating magnetic fields near the Larmor frequency of the precessing protons, energy can be transferred much faster, thus T_1 is short.

This can again be illustrated by our sandwich and race car example: (see page 17) handing over sandwiches (i.e. energy) from one car (proton) to the other (lattice) is easy and efficient, when both move with the same speed. With a difference in speeds, the energy transfer will be less efficient.

Why does fat have a short T_1 ?

The carbon bonds at the ends of the fatty acids have frequencies near the Larmor frequency, thus resulting in effective energy transfer.

And why is T_1 longer in stronger magnetic fields?

It is easy to imagine that in a stronger magnetic field it takes more energy for the protons to align against it. Thus these protons have more energy to hand down to the lattice, and this takes longer than handing down just a small amount of energy. Even though it may seem logical, this is the wrong explanation. As we heard in the beginning, the precession frequency depends on magnetic field strength, a relationship described by the Larmor equation. If we have a stronger magnetic field, then the protons precess faster. And when they precess faster, they have more problems handing down their energy to a lattice with more slowly fluctuating magnetic fields.

What influences T_2 ?

T_2 -relaxation comes about when protons get out of phase, which - as we already know - has two causes: inhomogeneities of the external magnetic field, and inhomogeneities of the local magnetic fields within the tissues (see page 29). As water molecules move around very fast, their local magnetic fields fluctuate fast, and thus kind of average each other out, so there are no big net differences in internal magnetic fields from place to place. And if there are no big differences in magnetic field strength inside of a tissue, the protons stay in step for a long time, and so T_2 is longer.

With impure liquids, e.g. those containing some larger molecules, there are bigger variations in the local magnetic fields. The larger molecules do not move around as fast, so their local magnetic fields do not cancel each other out as much. These larger differences in local magnetic fields consequently cause larger differences in precession frequencies, thus protons get out of phase faster, T_2 is shorter.

This can be illustrated by the following example: imagine that you drive down a street with many pot holes. When you drive slow (which is equal to the surroundings moving slow and

you standing still), you will be jumping up and down in your car with each pot hole. The differences in the surroundings (the magnetic field variations) influence you considerably. When you drive very fast (which is the same as if the surroundings move very fast), you do not feel the single pot holes anymore. Before they have a major effect on you, you are already back on the normal street level; thus their effect is averaged out, you are much less influenced by differences in the surroundings (the variations in magnetic field strength).

What does all this have to do with what we want to know? All these processes influence how your MR picture will finally look!



A brief review might be advisable

- T_1 is longer than T_2
- T_1 varies with the magnetic field strength; it is longer in stronger magnetic fields
- water has a long T_1 , fat has a short T_1
- T_2 of water is longer than the T_2 of impure liquids containing larger molecules.

Now let us perform an experiment

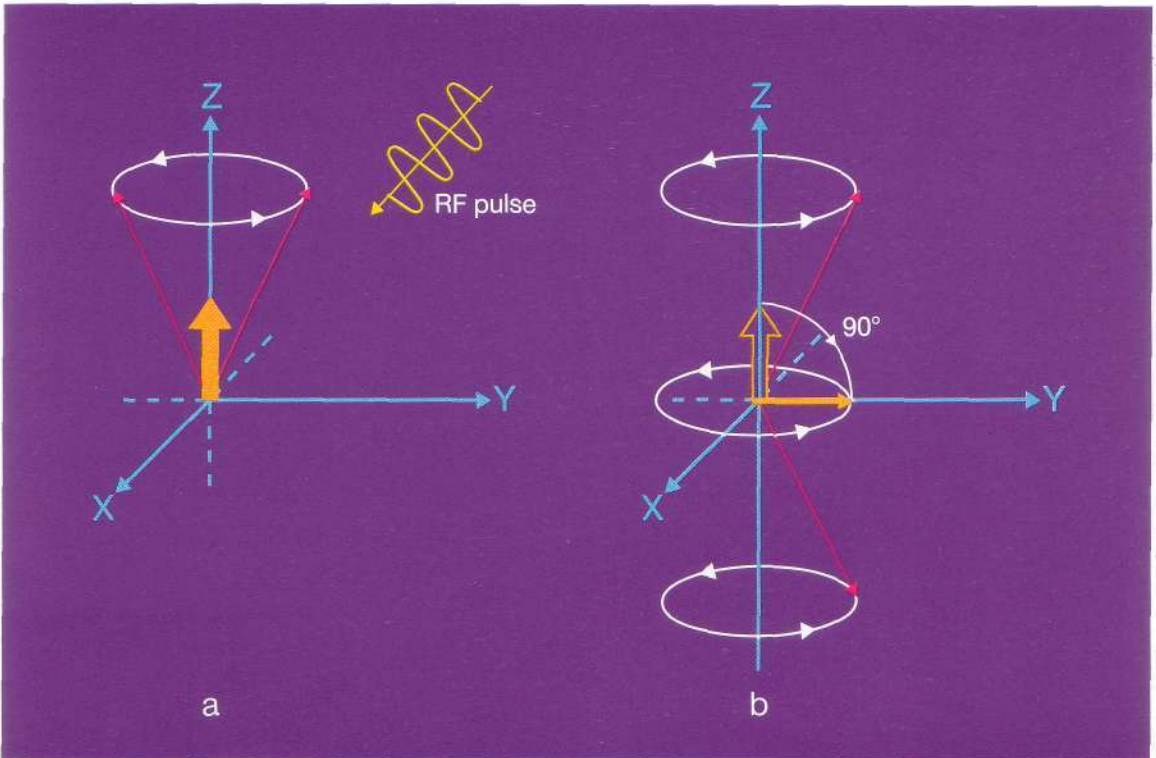
Look at figure 25, where you see two protons, precessing around the z-axis. I hope that you recall, that the z-axis indicates the direction of a magnetic field line (see page 11). Instead of only these two protons, in reality there may be 12 pointing up and 10 pointing down, or 102 up and 100 down - there shall only be two more protons pointing up. As we know, these are the ones that have a net magnetic effect because their effects are not cancelled out.

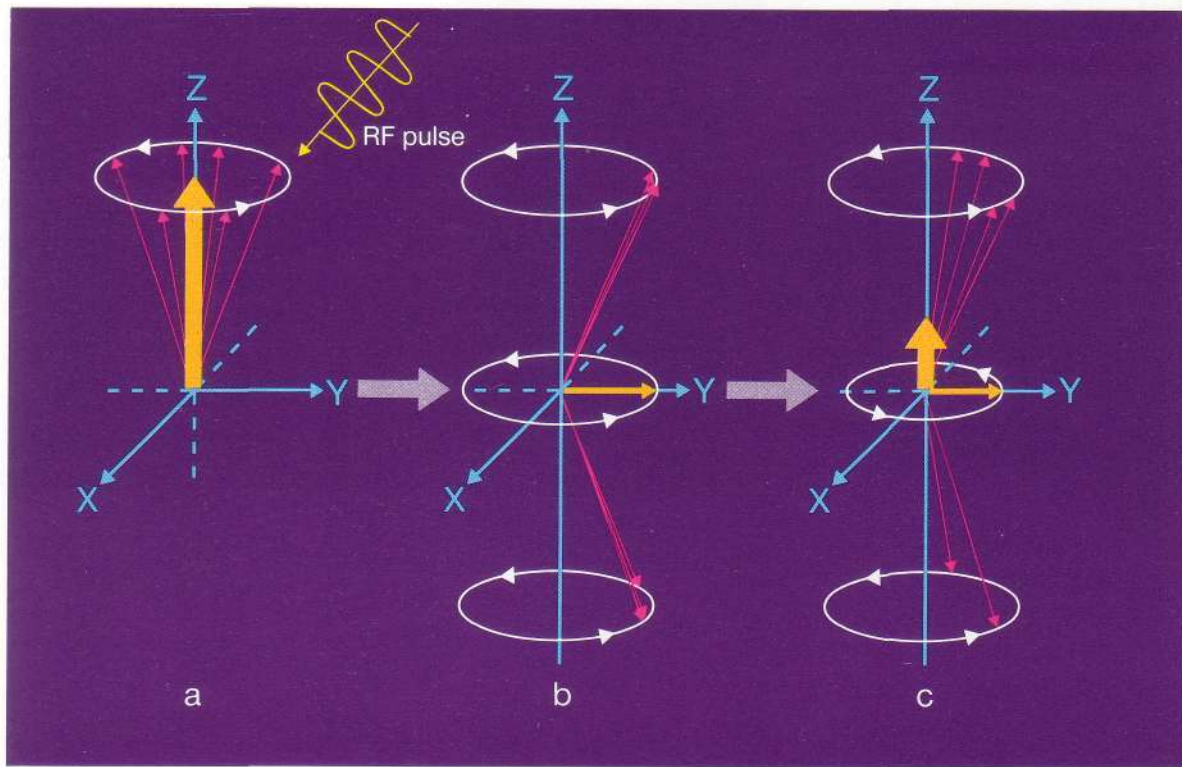
Now let us send in an RF pulse, which has just the correct strength and duration, that one of the two protons picks up energy, to go into the higher state of energy (points down/walks on its hands).

What will happen? The longitudinal magnetization (up to now resulting from two protons pointing up) will decrease, in our example to zero (one pointing up is neutralized by one pointing down). But: as both protons are in phase, now there is a transversal magnetization which had not been there before. This can be looked at - in effect - that a longitudinal magnetic vector is tilted 90° to the side (fig. 25).

An RF pulse which "tilts" the magnetization 90° is called a 90° pulse. Naturally, other RF pulses are also possible, and are named accordingly, e.g. 180° pulse.

Fig. 25
 If after the RF pulse, the number of protons on the higher energy level equals the number of protons on the lower energy level, longitudinal magnetization has disappeared, and there is only transversal magnetization due to phase coherence. The magnetic vector seems to have been "tilted" 90° to the side. The corresponding RF pulse is thus also called a 90° pulse.





To really understand this let us look at another example. In figure 26 (a) we have 6 protons pointing up; we send in a RF pulse, which lifts up 3 of them to a higher energy level (b).

The result: we no longer have a longitudinal, but a transversal magnetization (again having used a 90° pulse).

What happens, when the RF pulse is switched off?

Two things: protons go back to their lower state of energy, and they lose phase coherence.

It is important to note that both processes occur simultaneously and independently. For the sake of simplicity, let us look at what happens step by step, and first focus on the longitudinal magnetization:

- in 26 (c), one proton goes back to the lower energy state; resulting in 4 protons pointing up, and two pointing down. In net effect: we now have a longitudinal magnetization of "2".

- Then the next proton goes back up; now 5 protons point up, and one down, resulting in a net longitudinal magnetization of "4" (fig. 26d).

- After the next proton goes up, longitudinal magnetization equals "6" (fig. 26 e).

At the same time, transversal magnetization decreases (fig. 26c-e). Why? You should be able to answer this: the precessing protons lose phase coherence.

In fig. 26b, all protons point in the same direction, but then get increasingly out of phase, and thus kind of fan out (fig. 26c-e).

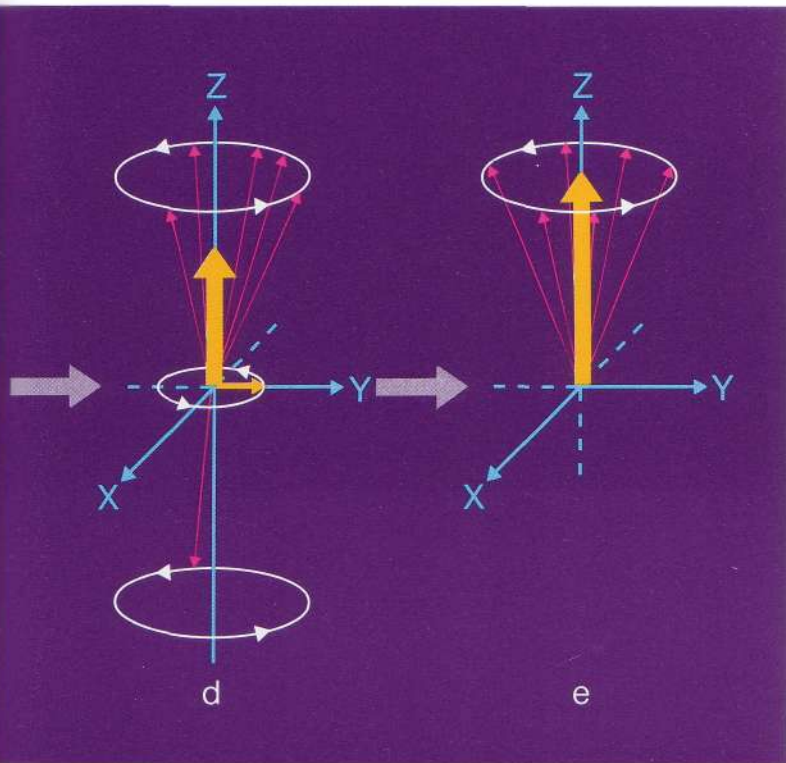
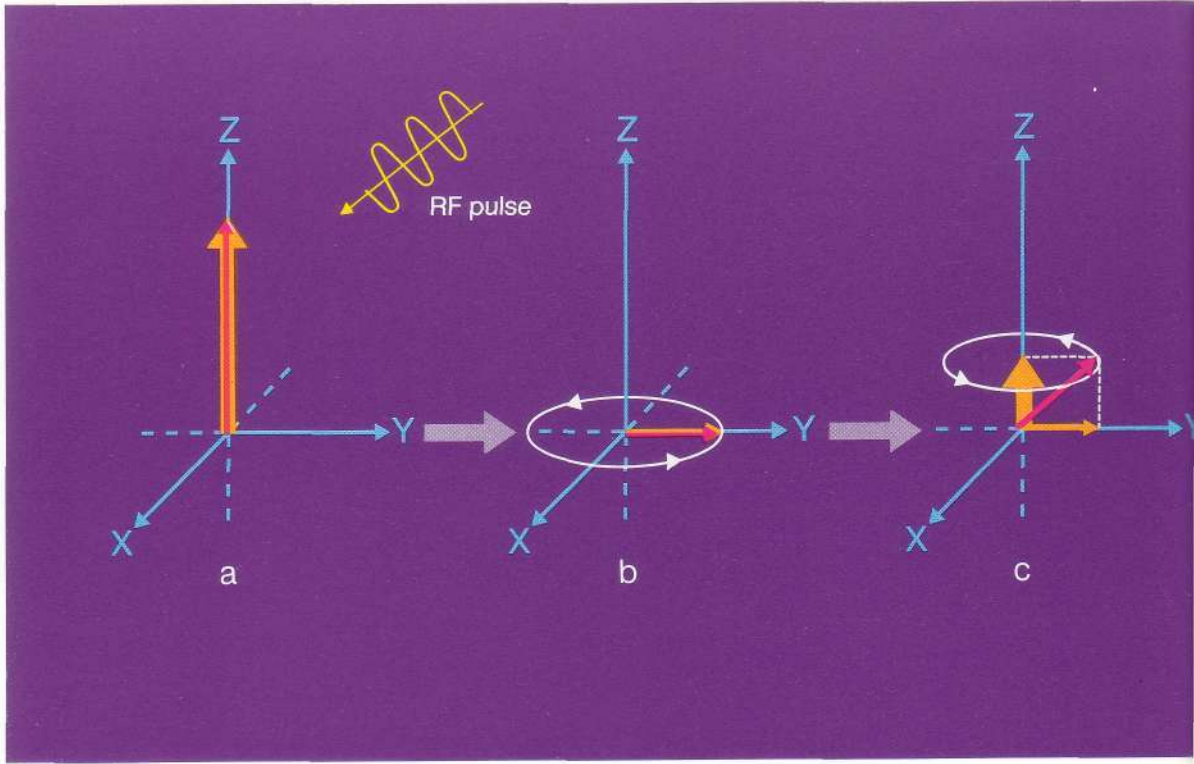


Fig. 26

In (a) is the situation before and in (b) immediately after an RF pulse is sent in. The RF pulse causes the longitudinal magnetization to decrease, and with a 90° pulse as illustrated, it becomes zero (b). Protons also start to precess in phase (b), which causes the new transversal magnetization. After the RF pulse is switched off (c-e) longitudinal magnetization increases, recovers, and transversal magnetization disappears, decays. Both processes are due to entirely different mechanisms, and occur independently even though at the same time.



In fig. 27, only the longitudinal and transversal magnetic vectors are depicted at corresponding times as in figure 26. These magnetic vectors add up to a sum vector. (—•)

If you have forgotten; vectors represent forces of a certain size and a certain direction. If you add vectors pointing to different directions up, you will come up with a direction that is somewhere in between, depending on the amount of force in the original directions.

This sum vector is very important, as it represents the total magnetic moment of a tissue in general, and thus can be used instead of the two single vectors, representing longitudinal and transversal magnetization separately. Our magnetic sum vector during relaxation goes back to a longitudinal direction, in the end equaling the longitudinal magnetization.

What we have to remember is that this whole system actually is precessing, including the sum magnetic vector/moment. And thus the sum vector will actually perform a spiraling motion (fig. 27f).

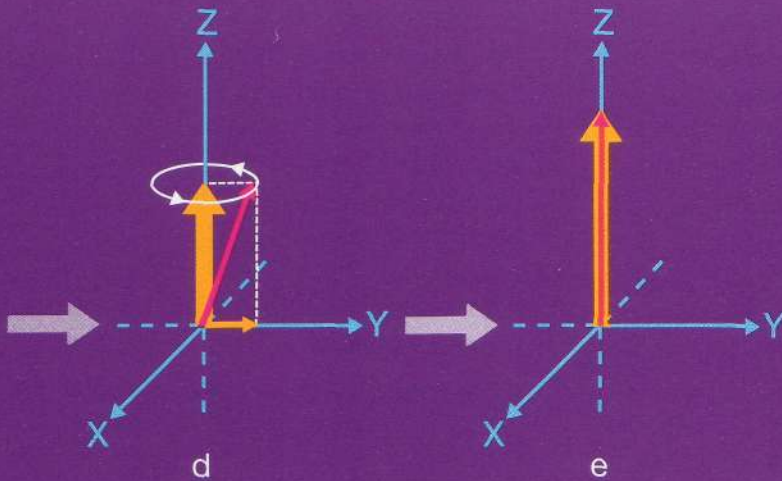
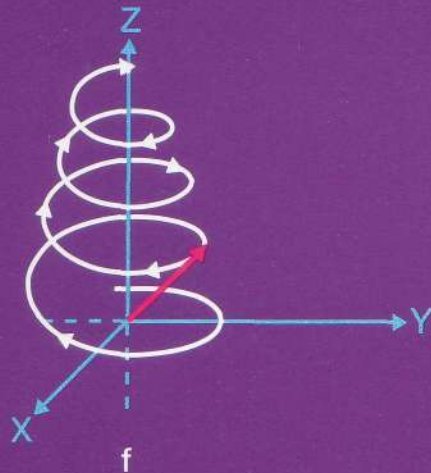


Fig. 27

In this illustration only the longitudinal and transversal magnetization vectors from our experiment in figure 26 are depicted. In (a) - before the RF pulse - there is only longitudinal magnetization. Immediately after the 90° RF pulse there is no longitudinal but new transversal magnetization (b), and this transversal magnetization vector is spinning around. With time this transversal magnetization decreases, while longitudinal magnetization increases (c-d), until the starting point with no transversal but full longitudinal magnetization is reached again (e). Transversal and longitudinal magnetization vectors add up to a sum vector (\bullet). This sum vector performs a spiraling motion (f) when it changes its direction from being in the transversal (x-y) plane (no longitudinal magnetization) to its final position along the z-axis (no transversal magnetization).



I hope that you recall, that a changing magnetic force/moment can induce an electrical current, which was the signal that we receive and use in MR. If we put up an antenna somewhere, as in figure 28, we will get a signal as illustrated.

This is easy to imagine, if you think of the antenna as a microphone, and the sum magnetic vector as having a bell at its tip. The further the vector goes away from the microphone, the less loud the sound. The frequency of the sound, however, remains the same because the sum vector spins with the precessing frequency (fig. 29).

This type of signal is called a FID signal, from free induction decay. It is easy to imagine, that you get a very good strong signal directly after the 90° RF pulse (as the bell comes very close to the microphone in our example).

By now it should be obvious that the magnetic vectors directly determine the MRI signal and signal intensity by inducing electrical currents in the antenna. Instead of the terms "longitudinal" or "transversal magnetization", we can also use the term "signal or signal intensity" at the axis of our T_1 , and T_2 curves. This will hopefully become clearer as you continue reading.

Fig. 28

For an external observer, the sum vector of figure 27f constantly changes its direction and magnitude while it performs its spiraling motion. The sum vector induces an electrical current in an antenna, the MR signal. This is of greatest magnitude immediately after the RF pulse is switched off and then decreases.

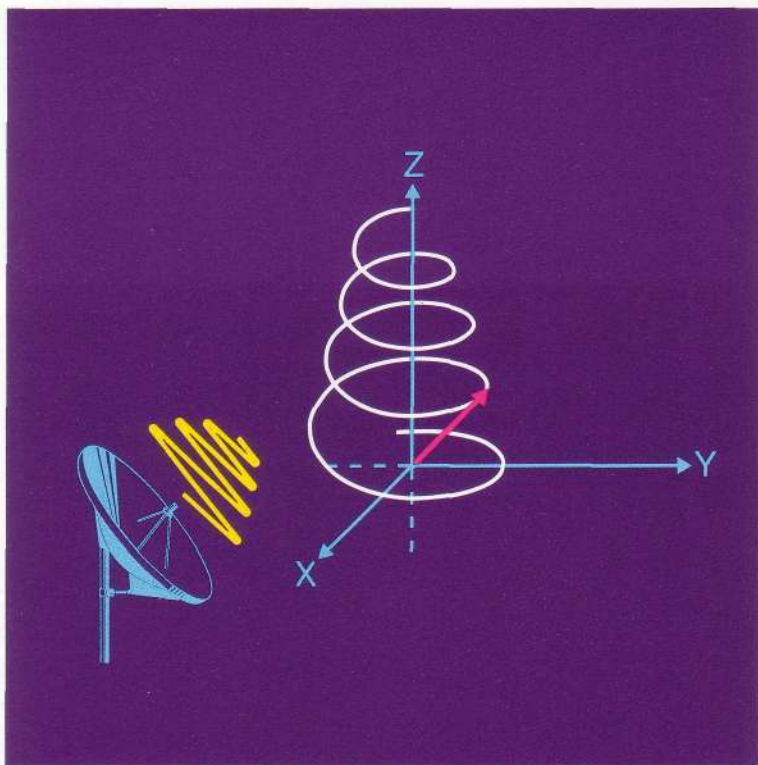
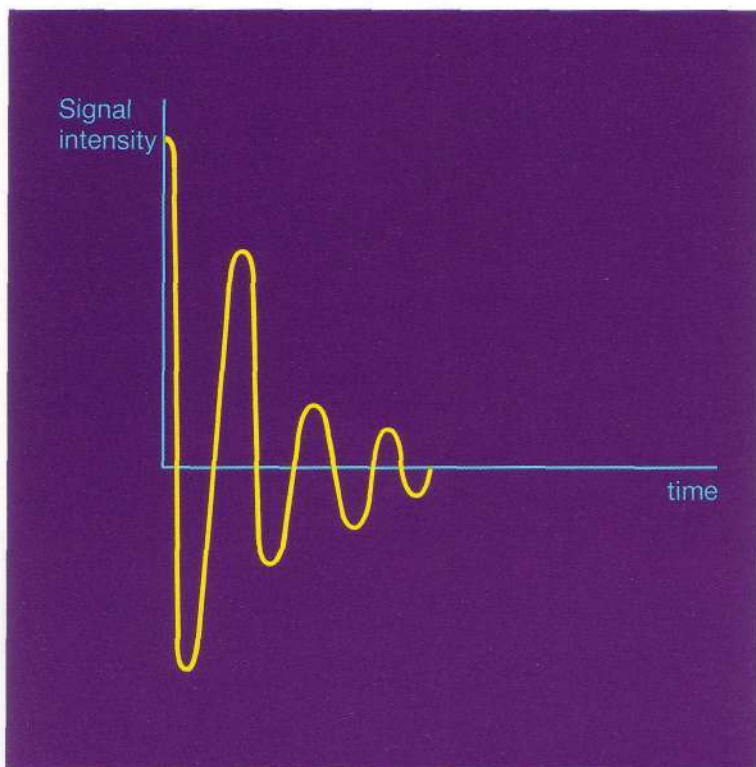


Fig. 29

The signal from our experiment in figures 26 to 28 disappears with time, however, has a constant frequency. This type of signal is called a FID (free induction decay) signal.



What about another experiment?

Let us perform another experiment similar to the one above, which is illustrated in figure 30. In figure 30a, we have two tissues, A and B, which have different relaxation times (tissue A has a shorter transversal as well as longitudinal relaxation time). We send in a 90° RF pulse, and wait a certain time TR_{long} (we will explain later why we use the term TR), and then send in a second 90° pulse. What will happen?

As after the time TR_{long} tissue A and tissue B have regained all of their longitudinal magnetization (frame 5), the transversal magnetization after the second pulse will be the same for both tissues, as it was in frame 1.

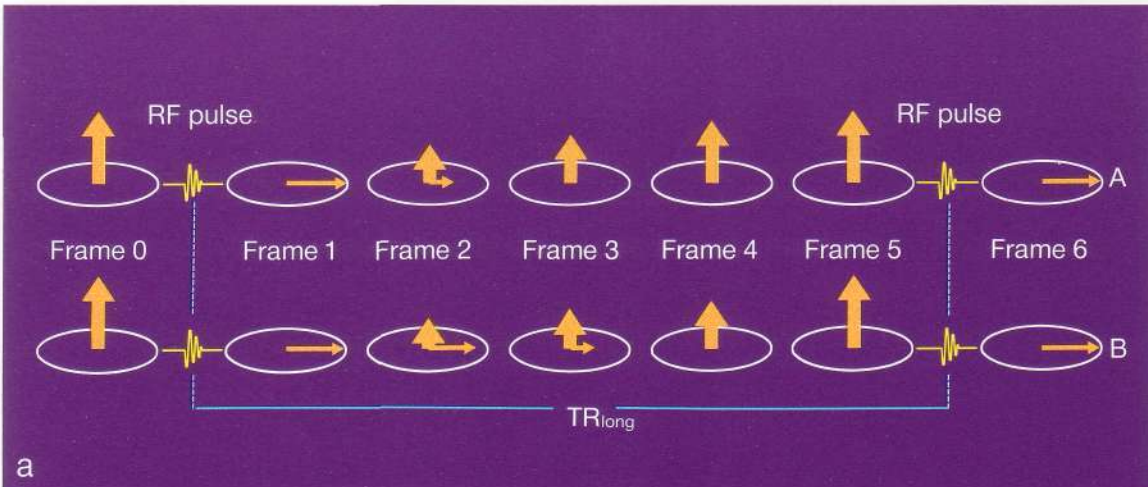
What if we do not wait so long from pulse to pulse? Look at figure 30b, where the second 90° pulse is sent in after the time TR_{short} , i.e. after frame 4 already. At this time, tissue A has regained more of its longitudinal magnetization than tissue B. When the second 90° pulse now "tilts" the longitudinal magnetization 90 degrees, the transversal magnetic vector of tissue A is larger than that of tissue B. And when this vector of A is larger, it will reach closer to our antenna; thus the imaginary bell at the tip of vector A will cause a louder, stronger signal in our "microphone", the antenna, than vector B.

The difference in signal intensity in this experiment depends on the difference in longitudinal magnetization, and this means on the difference in T_1 between the tissues.

Using these two pulses, we can now differentiate tissue A from tissue B, which might have been impossible choosing only one 90° pulse or two 90° pulses that are a long time apart (after a long time, the differences in T_1 between tissue A and B no longer play a role in our experiment, because after that time the tissue B with the longer T_1 is back to its original state, too).

Fig. 30a

A and B are two tissues with different relaxation times. Frame 0 shows the situation before, frame 1 immediately after a 90° pulse. When we wait for a long time (TR_{long}) the longitudinal magnetization of both tissues will have totally recovered (frame 5). A second 90° pulse after this time results in the same amount of transversal magnetization (frame 6) for both tissues, as was observed after the first RF pulse (frame 1).



When you use more than one RF pulse - a succession of RF pulses - you use a so-called pulse sequence. As you can use different pulses, e.g. 90° or 180° pulses, and the time intervals between successive pulses can be different, there can be many different pulse sequences. As we saw in our experiment, the choice of a pulse sequence will determine what kind of signal you get out of a tissue. So it is necessary to carefully choose and also describe the pulse sequence for a specific study.

The pulse sequence that we used was made up of one type of pulse only, the 90° pulse. This was repeated after a certain time, which is called

$TR = \text{time to repeat}$

How did TR influence the signal in our experiment?

With a long TR we got similar signals from both tissues, both

would appear the same on a MR picture. Using a shorter TR, there was a difference in signal intensity between the tissues, determined by their difference in T_1 . The resulting picture is called a T_1 -weighted picture. This means, that the difference of signal intensity between tissues in that picture, the tissue contrast, is mainly due to their difference in T_1 . However, there is always more than one parameter influencing the tissue contrast; in our example, T_1 is just the most outstanding one.

What is a short or a long TR?

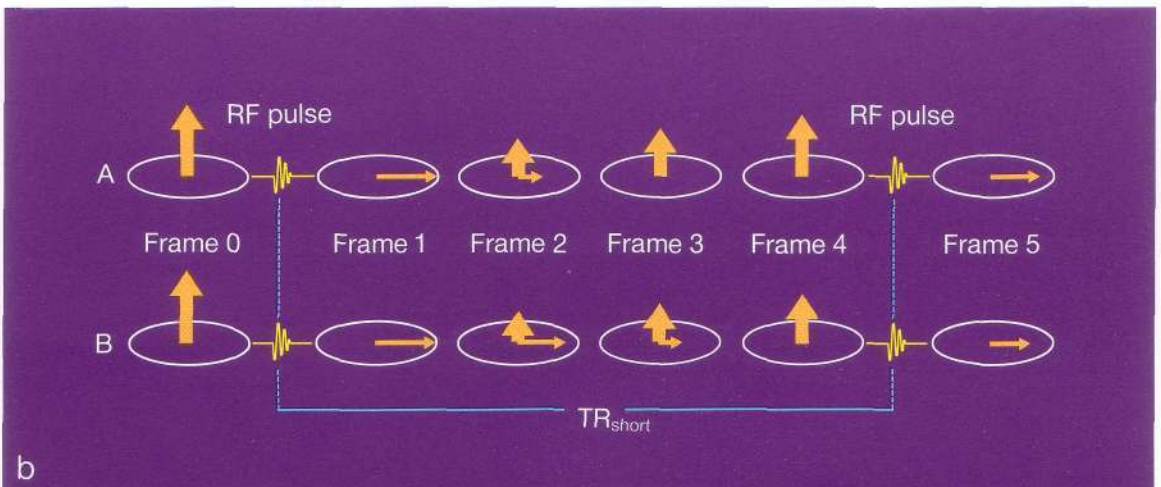
A TR of less than 500 msec is considered to be short, a TR greater than 1500 msec to be long.

As you may imagine or know already, we can also produce T_2 -weighted images, and so-called proton density (-weighted) images.

That proton density, which is also called spin density, influences tissue contrast, can be explained quite simply; where there are no protons, there will be no signal; where there are many protons, we will have "lots" of signals. We will read more about this later. The point is that by using certain pulse sequences, we can make certain tissue characteristics to be more or less important in the resulting image.

Fig. 30b

When we do not wait as long as in figure 30a, but send in the second RF pulse after a shorter time (TR_{short}), longitudinal magnetization of tissue B, which has the longer T_1 , has not recovered as much as that of tissue A with the shorter T_1 . The transversal magnetization of the two tissues after the second RF pulse will then be different (frame 5). Thus, by changing the time between successive RF pulses, we can influence and modify magnetization and the signal intensity of tissues.



By choosing a pulse sequence, the doctor can be compared to a conductor of an orchestra (figure 31): he can influence the overall appearance of the sound (signal), by making certain instruments (parameters) influence the sound more than others. All instruments (parameters), however, always play some role in the final sound (signal).

Fig. 31

The MRI doctor can be compared to a conductor: by choosing certain pulse sequences, he can modify the resulting signal, which is itself influenced by different parameters.



Let us go back to our experiment once more for a short repetition

- With a certain type of RF pulse we can cause the longitudinal magnetization to disappear, while a transversal magnetization appears. The "net magnetization" (the sum vector of longitudinal and transversal magnetization) is "tilted" 90° in this case (when we started we only had longitudinal magnetization). The corresponding RF pulse is therefore called a 90° pulse.
- The transversal component of the net magnetization can induce a measurable signal in an antenna
- Immediately after the RF pulse relaxation begins; transversal magnetization starts to disappear and longitudinal relaxation begins to reappear. The sum magnetic vector goes back to its original longitudinal alignment, the signal disappears.

- When we send in the second 90° pulse, the net magnetization is again tilted 90° , and we again receive a signal.
- The strength of this signal depends (among other things) on the amount of longitudinal magnetization we start out with. Do you remember the T_1 -curve (if not see page 27)? The T_1 -curve described the relationship between time (after an RF pulse) and the amount of longitudinal magnetization (fig. 18). When we wait a long time until we send in our second RF pulse, longitudinal magnetization will have recovered totally. The signal after the second RF pulse will thus be the same as the one after the first pulse. However, when the second pulse comes in earlier, the signal will be different, as the amount of longitudinal magnetization at that time is less.

In fig. 32 we plotted the T_1 -curves for brain and for cerebrospinal fluid (CSF). At the time 0 we have no longitudinal magnetization at all, and this can be the time immediately after our first 90° pulse. When we wait a long time before we repeat the 90° pulse (TR_{long}), longitudinal magnetization has pretty much recovered. The longitudinal magnetic vectors that will be "tilted" 90° differ only in a small amount, so there will be only a small difference in signal intensity, i.e. tissue contrast between brain and CSF. If we, however, send in the second pulse after the shorter TR_{short} , the difference in longitudinal magnetization is rather large, so there will be a better tissue contrast. And as we can see from the distance between the two curves, there is a time span where tissue contrast is most pronounced.

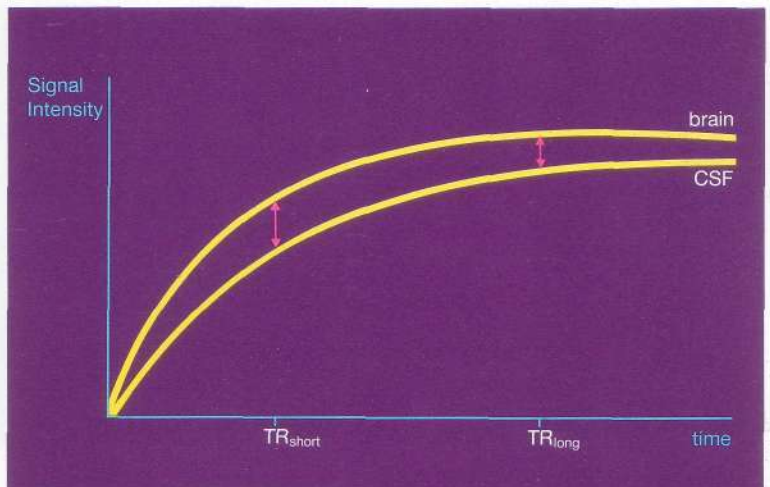


Fig. 32
Brain has a shorter longitudinal relaxation time than CSF. With a short TR the signal intensities of brain and CSF differ more than after a long TR.

Why are the signals after a very long time TR between pulses not identical?

We have heard the explanation already. The signal intensity depends on many parameters. When we wait a long time, T_1 does not influence the tissue contrast any more, however, there may still be a possible difference in the proton density of the tissues in question. And when we wait a very long time TR in our experiment from fig. 32, the difference in signal is mainly due to different proton densities, we have a so-called proton density (or spin density) weighted image.

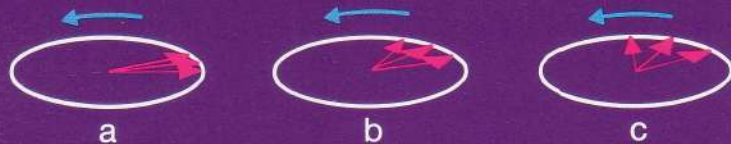
Now we have read about T_1 - and proton density-weighted images.

How do we obtain a T_2 -weighted image?

This is a little more difficult to understand. Let us perform another experiment, which is a little different from the ones before. First, we use a 90° pulse. The longitudinal magnetization is tilted, we get a transversal magnetization. What happens after this pulse, when we wait a short time? You should be able to answer this question without difficulties - if not go back to page 37 before you continue to read. After the pulse is switched off, longitudinal magnetization starts to reappear, the transversal

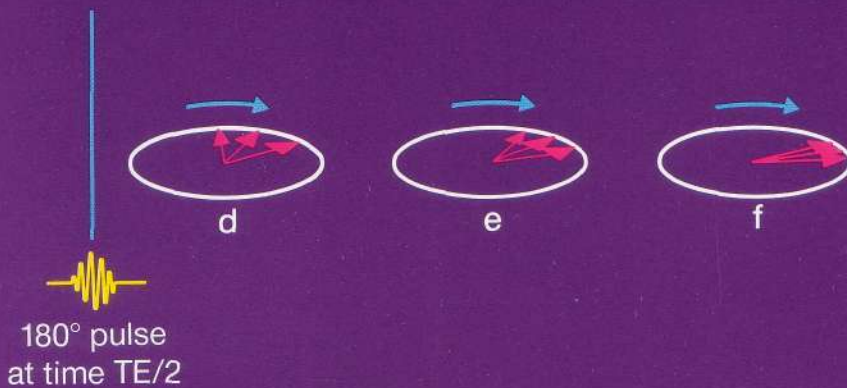
magnetization, however, starts to disappear. Why does the transversal magnetization disappear? The protons lose phase coherence, as we have heard above. And this is illustrated in figure 33 for three protons, which are almost exactly in phase in (a) but increasingly spread out, as they have different precession frequencies (b and c). The loss of phase coherence results in decreasing transversal magnetization and thus loss of signal.

Now we do something new: after a certain time (which we call $TE/2$, half of TE , for reasons you will understand in a few minutes) we send in a 180° pulse. What does this do?



The 180° pulse acts like a rubber wall; it makes the protons turn around, so that they precess in exactly the opposite direction, which is clockwise in figure 33d. The result is that the faster precessing protons are now behind the slower ones. If we wait another time $TE/2$, the faster ones will have caught up with the slower ones (fig. 33f). At that time the protons are nearly in phase again, which results in a stronger transverse magnetization, and thus in a stronger signal again. A little later however, the faster precessing protons will be ahead again, with the signal decreasing again.

Fig. 33
After the RF pulse is switched off, the protons dephase (a-c). The 180° pulse causes them to precess in the opposite direction and so they rephase again (d-f).



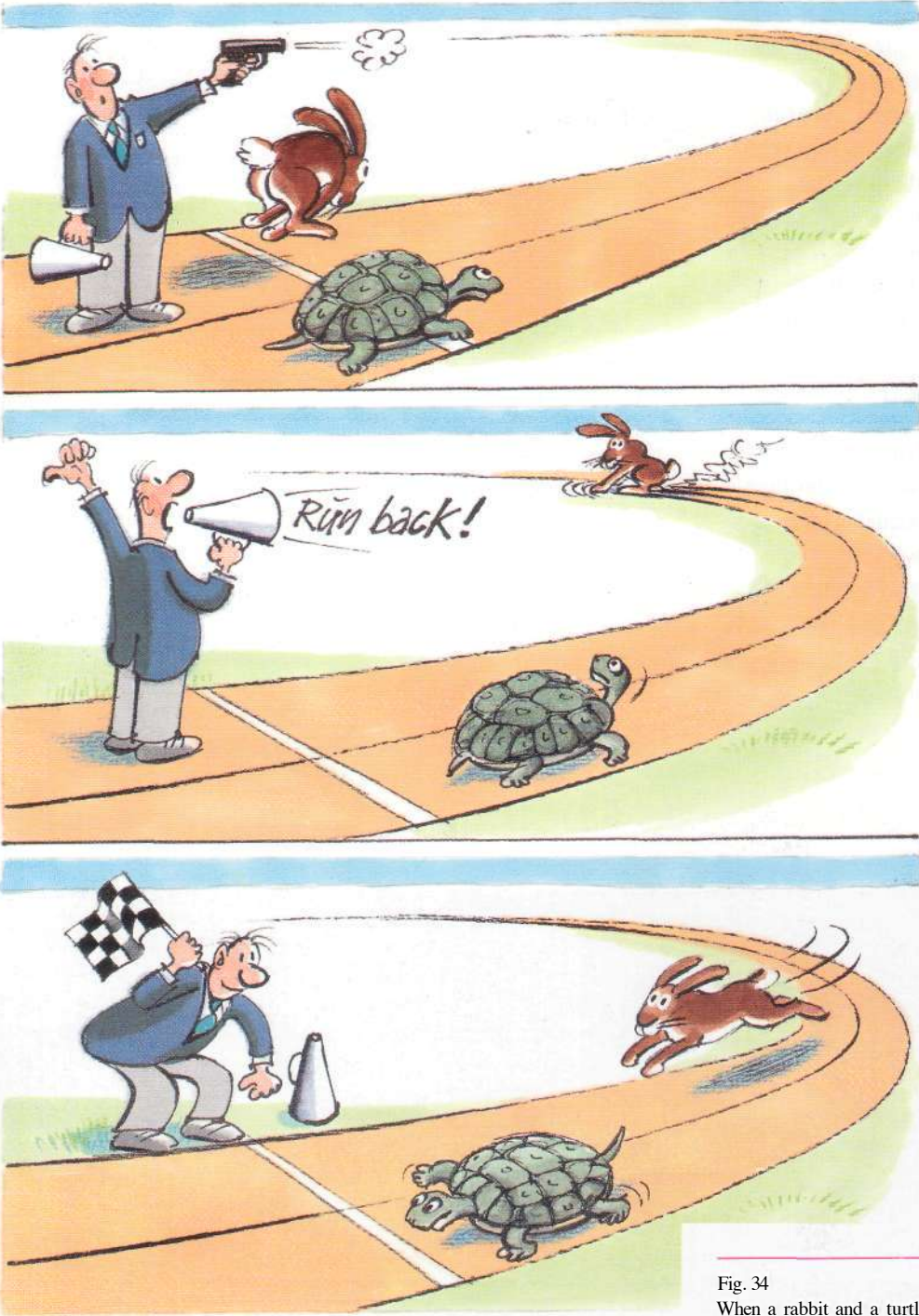


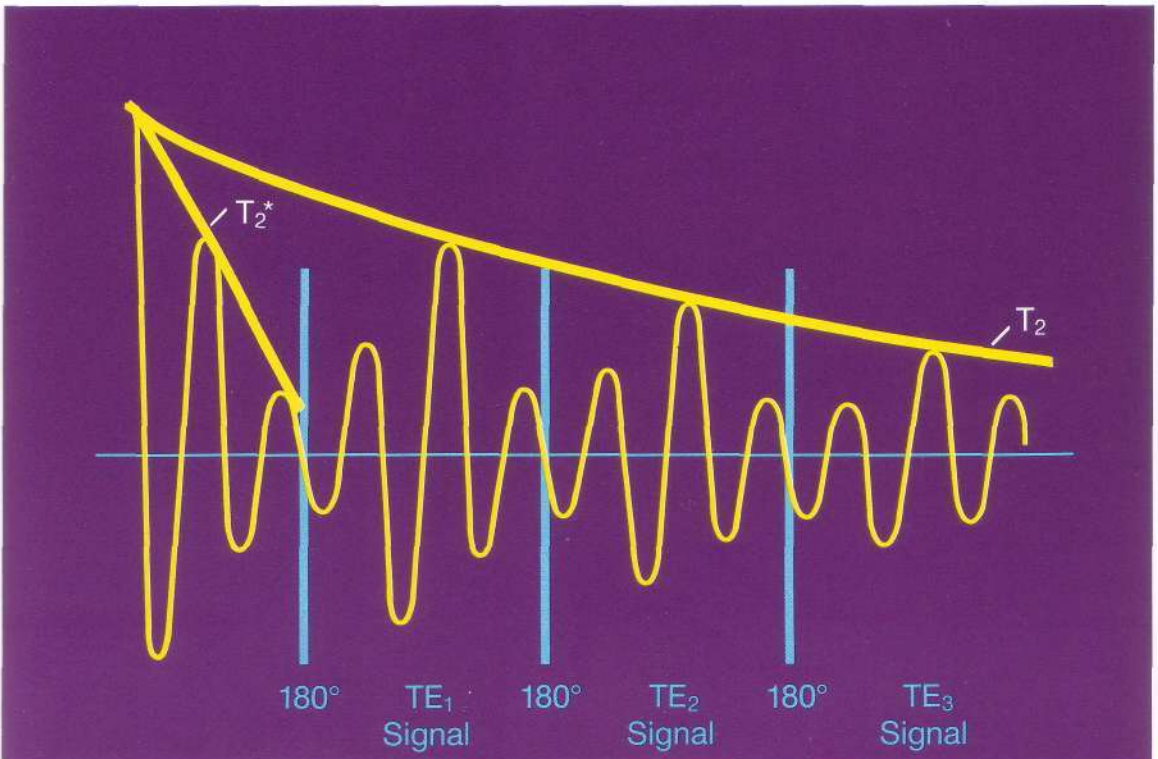
Fig. 34
When a rabbit and a turtle run in one direction for a certain time, then turn around and run in the opposite direction with the same speed for the same time, they will arrive at the starting point at the same time.

To illustrate this: think about a race between a rabbit and a turtle starting at the same line (fig. 34). After a certain time ($TE/2$), the rabbit is ahead of the turtle. When you make the competitors run in the opposite direction for the same length of time, they will both be back at the starting line at exactly the same time (assuming, that they run with constant speed).

The 180° pulse in our experiment acts like a wall, from which the protons bounce back, like a mountain reflecting sound waves as echoes. This is why the resulting strong signal is also called an echo, or spin echo.

After we have our signal, our spin echo, the protons lose phase coherence again, the faster ones getting ahead as we have heard. We naturally can perform the experiment again with another 180° pulse, and another and another . . . If we now plot time vs. signal intensity, we get a curve like in fig. 35.

Fig. 35
The 180° pulse refocusses the dephasing protons which results in a stronger signal, the spin echo after the time TE. The protons then dephase again and can be refocussed another time by a 180° pulse and so on. Thus it is possible to obtain more than one signal, more than one spin echo. The spin echoes, however, differ in intensity due to so-called T_2 -effects.
A curve connecting the spin echo intensities is the T_2 curve. If we did not use the 180° pulse, the signal intensity would decay much faster. A curve describing the signal intensity in that case is the T_2^* (T_2 star) curve, which is described in a little more detail on page 54.



From this curve we can see, that the spin echo, the resulting signal, decreases with time. Responsible for this is the fact that our 180° pulse only "neutralizes" effects that influence the protons in a constant manner, and these are the constant inhomogeneities of the external magnetic field. Inconstant inhomogeneities from local magnetic fields inside of the tissue cannot be "evened out", as they may influence some protons before the 180° pulse differently than after the 180° pulse. So some of the protons may still be behind or in front of the majority of the protons, that will reach the starting line at the same time. So from echo to echo the intensity of the signal goes down due to the so-called T_2 -effects.

Maybe we should illustrate this by an example: imagine two buses full of people, e.g. after a soccer or football game. They are standing at a starting line (fig. 36). With two microphones, you record the signal (e.g. the singing from the crowd) that is coming from each bus. Then the buses leave in the same direction. Recording the signal, you may recognize, that one signal disappears faster, than the other.

This can have two different causes: may be one bus drives faster than the other (loss of signal would thus be due to external influences, the external magnetic field inhomogeneities).

Or the difference in signal intensities, the difference in singing, may be due to differences of inherent properties of the two groups (internal inhomogeneities); may be in one bus, there are only the "party animals", who do not become tired as fast, as the people in the other crowd.

To figure out what actually is the reason for the signal disappearing, you can make the buses turn around after a certain time $TE/2$, and have them drive back with the same speed also for the time $TE/2$. After time $2 \times TE/2 = TE$ the buses will be back at the starting line. The signal intensity, that you record with your microphone then depends only on inherent properties, i.e. how tired the crowds have become.

If you do not use a 180° pulse to neutralize constant external inhomogeneities, the protons will experience larger differences in magnetic field strength when the RF pulse is switched off. Due to this they will be out of phase faster, the transversal relaxation time will be shorter. To distinguish this shorter transversal relaxation time from the T_2 after the 180°

pulse, it is called T_2^* (T_2 star). The corresponding effects are named T_2^* -effects. These T_2^* -effects are important with the so-called fast imaging sequences (see page 81).

In our example with the buses this would mean that we just record the signals as the buses drive away. The signals vanish due to extrinsic (bus speed) and intrinsic (exhaustion of the passengers) properties under these circumstances (see fig. 36).

The type of pulse sequence, that we used in our experiment, is called a spin echo sequence, consisting of a 90° pulse and a 180° pulse (causing the echo). This pulse sequence is very important in MR imaging, as it is the workhorse among the pulse sequences, which can be used for many things. It is important to realize that with a spin echo sequence, we cannot only produce T_2 -, but also T_1 - and proton density - weighted pictures. We will get to that a little later.

Fig. 36

Without having the buses come back (i.e. a 180° pulse), it is impossible to say whether a decrease in signal intensity is due to inherent tissue properties (the differing shape of the bus passengers), or due to external influences, i.e. different bus speeds.



Let us first look at such a T_2 -weighted sequence

What did we do? First we sent in a 90° pulse, resulting in some transversal magnetization. Immediately after the 90° pulse we have a maximum transversal magnetization. However, this transversal magnetization disappears, due to T_2 -effects. How fast transversal magnetization disappears, can be seen from a T_2 -curve; in fig. 37 we have plotted T_2 -curves for two different tissues, tissue A having a short T_2 (e.g. brain), tissue B having a long T_2 (water or CSF). Both curves start at 0, which is

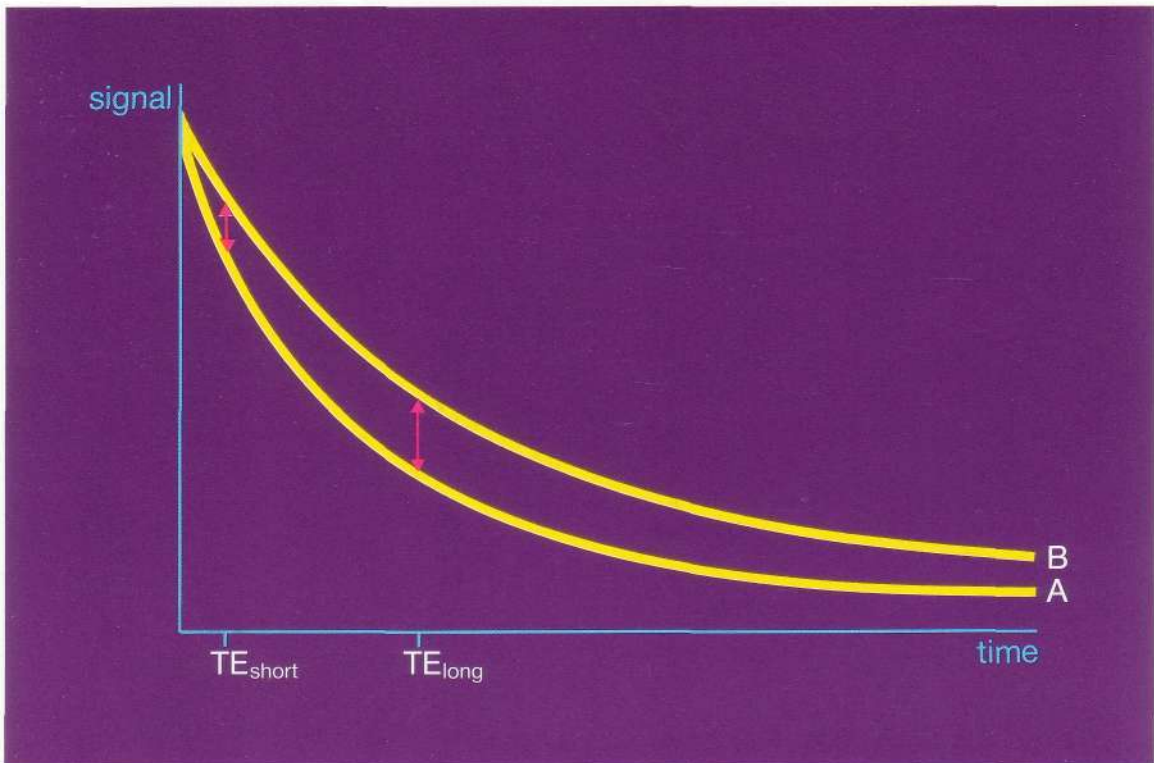
the time immediately after the 90° pulse is switched off. When we wait for a certain time $TE/2$ to send in the 180° pulse, transversal magnetization will have become smaller. After waiting another time $TE/2$ (that is TE after the 90° pulse is switched off), we will receive a signal, the spin echo. The intensity of this echo is given by the T_2 -curve at the time TE . This time TE between the 90° pulse and the spin echo is called

$$\underline{TE} = \text{time to echo.}$$

The time TE can be chosen by the operator. And as we can see from the T_2 -curves, TE influences the resulting signal, and thus also the image:

Fig. 37

T_2 -curves for two tissues with different transversal relaxation times; tissue A has a shorter T_2 than tissue B, thus loses transversal magnetization faster. With a short TE (TE_{short}) the difference in signal intensity is less pronounced than after a longer TE (TE_{long}).



the shorter the time TE , the stronger the signal that we get from a tissue.
To get the best, strong signal, it may seem reasonable to use a short TE , because with longer TE s signal intensity decreases. With a short TE , however, there will be a problem (fig. 37). Both T_2 -curves in this example start at the same point. If we only wait a very short TE , the difference in signal intensity between tissue A and tissue B is very small, both tissues may hardly be distinguished as there is hardly any contrast (which is the difference in signal intensity of tissues). Consequence: with a short TE , differences in T_2 do not influence tissue contrast

very much. As both T_2 -curves diverge, with a longer TE the difference in T_2 -curves, and thus the difference in signal intensity = contrast, is more pronounced in our example. So it might be reasonable to wait a very long TE ; the resulting picture should be very heavily T_2 -weighted. But (and there is always a "but") if we wait longer, the total signal intensity becomes smaller and smaller. The signal to noise ratio becomes smaller, the picture appears grainy. An example to illustrate this signal-to-noise problem: when you receive a local radio station in your radio, this gives you a good signal, e.g. loud music

and only little static noise. When you drive out of town, the signal intensity of the radio station becomes weaker, and you will hear more static noise; and when you drive even further away, you may not be able to discern the music from the background noise. And this is the same for the MR signal: there is always some noise in the system, however, when the signal is strong, this does not matter much. However, the smaller the signal, the harder it is, to differentiate it from the background noise.



Let us review some facts

We have learned that

- the spin echo sequence consists of a 90° and a 180° pulse
- after the 90° pulse protons are dephasing due to external and internal magnetic field inhomogeneities
- the 180° pulse rephases the dephasing protons (sometimes the term spins is used interchangeable for protons), and a stronger signal, the spin echo results
- the 180° pulse serves to "neutralize" the external magnetic field inhomogeneities
- signal decrease from one echo to the next, when using multiple 180° pulses is due to internal T_2 -effects
- by choosing different TEs (different times after the 90° pulse) the signals can be T_2 -weighted in varying degrees - with very short TEs, T_2 -effects have not yet had time to really show up

with longer TEs, the signal intensity difference between tissues will be depending very much on their T_2 s, their transverse relaxation times

- with very long TEs, there should be even more T_2 -weighting, however, signal intensity as such would be so small, that at best it can just barely be distinguished from the background noise

What actually is a short/long TR or TE?

A short TR is one that is about as short as the smallest/shortest T_1 that we are interested in (Remember: T_1 was a time constant, not the time that it takes for a tissue to regain its longitudinal magnetization!). A long TR is about 3 times as long as a short TR.

A TR of less than 500 msec is considered to be short, a TR of more than 1500 msec to be long (just to give you a rough idea).

A short TE is one that is as short as possible, a long TE also is about 3 times as long.

A TE of less than 30 msec is considered to be short, a TE greater than 80 msec to be long.

Let us go back to our spin echo pulse sequence

This sequence can be illustrated schematically as in figure 38:

90° pulse - wait TE/2 - 180° pulse - wait TE/2 - record signal

For certain different reasons, such a pulse sequence is repeated two or more times. The time to repeat a pulse sequence was TR, time to repeat; so what we get is the following scheme:

1. (90° - TE/2 - 180° - TE/2 -> record signal at TE)

after TR (time from the beginning of one 90° pulse to the next 90° pulse) follows another pulse cycle and signal measurement:

2. (90° - TE/2 - 180° - TE/2 -> record signal at TE)

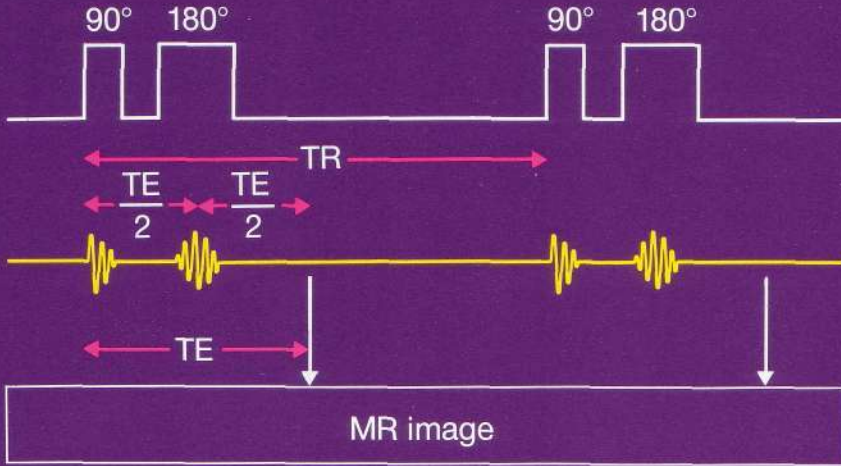
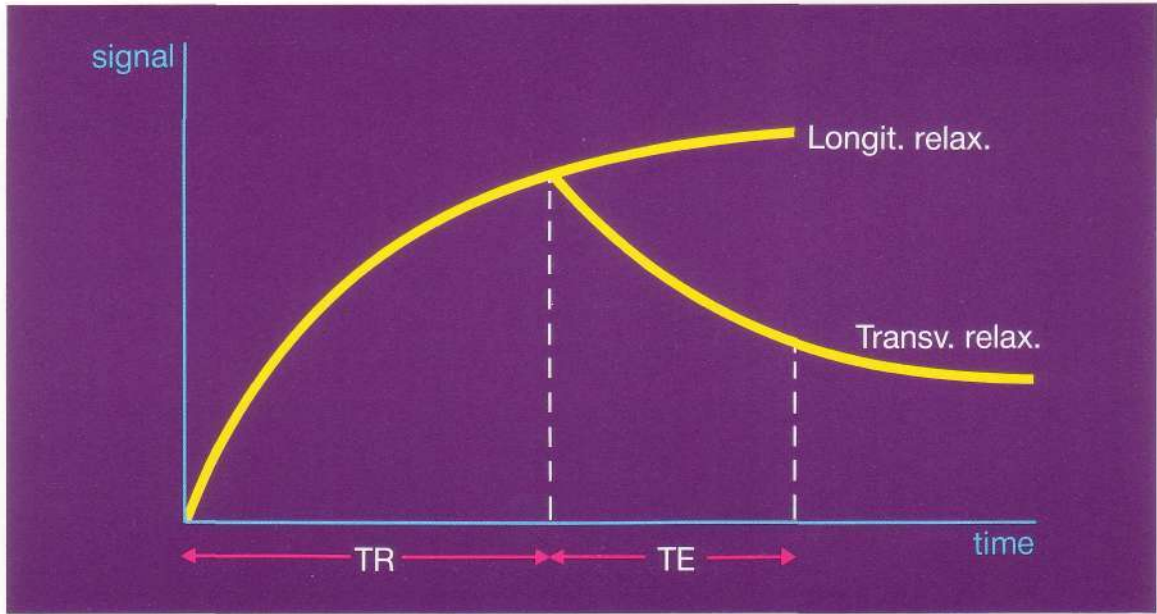


Fig. 38
Schematic illustration of a spin echo pulse sequence.



To figure out how much signal you get from a certain tissue with certain parameters of a spin echo sequence, you actually have to do no more than combine its T_1 - and T_2 -curves, as is illustrated in figure 39. There we have the T_1 and T_2 -curve of a certain tissue.

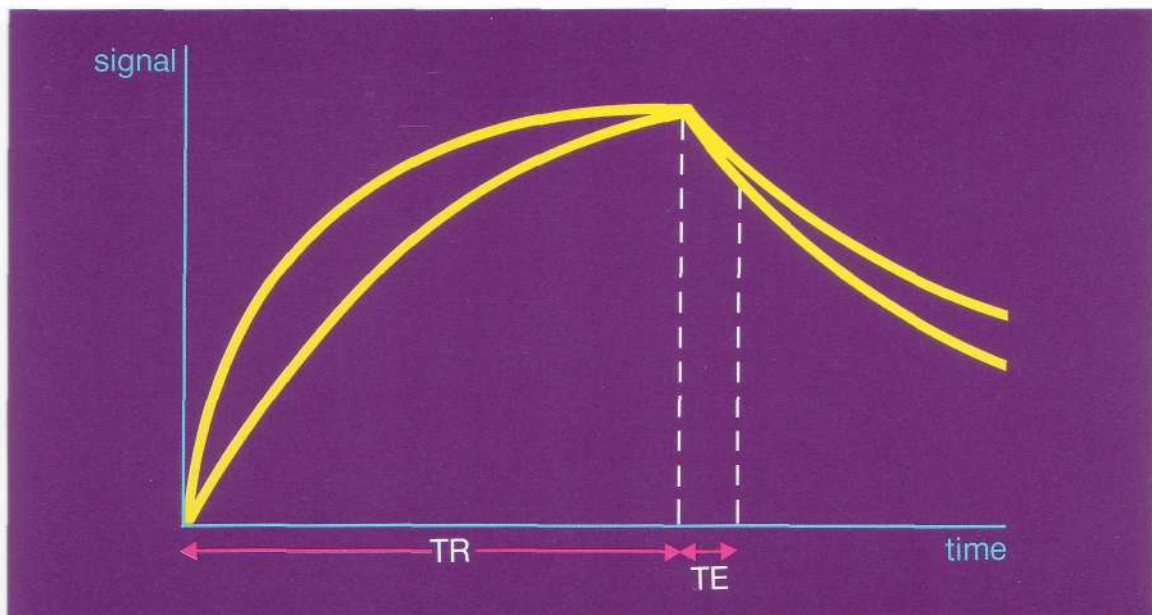
Which parameter determined the amount of longitudinal magnetization?

That was TR. To see how much longitudinal magnetization will be tilted 90° to the side (and thus to figure out, with how much transversal magnetization we start out with), we just look at the intensity of the longitudinal magnetization at the time TR. The longitudinal magnetization at this point, "tilted" in the transversal plane, is

the starting point from which transversal magnetization decays. So we just attach the T_2 -curve at this point. How much signal we get with a spin echo sequence to construct the image, also depends on TE, the time that we wait after the 90° pulse. So we now only have to look for the signal intensity at the time TE on the T_2 -curve.

Fig. 39

It is possible to determine signal intensity for a tissue using a spin echo sequence by combining the T_1 - and the T_2 -curve for that tissue. The longitudinal magnetization after the time TR is equal to the amount of transversal magnetization we start out with, as it is "tilted" 90° degrees. This transversal magnetization immediately starts to disappear by a rate which is determined by the transversal relaxation time, and thus by the T_2 -curve. The signal intensity of the tissue after a time TE can then be inferred from the T_2 -curve at this time TE (which starts after TR!).



What picture do we get, when we choose a long TR and a short TE?

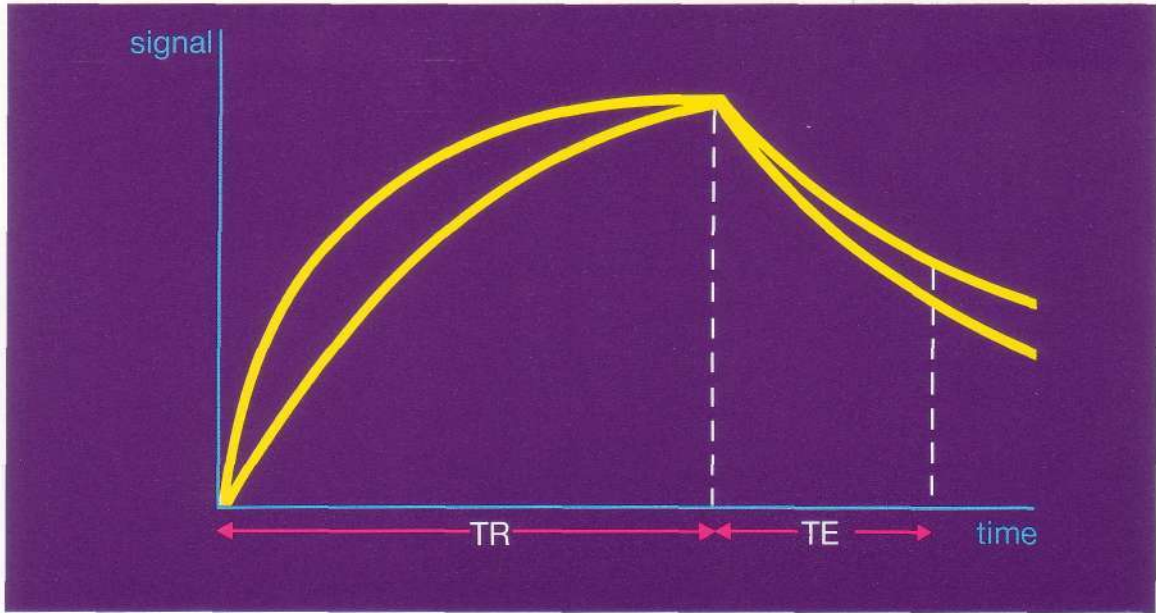
This is illustrated in figure 40, where the T_1 - and T_2 -curves for two different tissues are depicted. Maybe you remember our experiment from page 46, when we sent in a 90° pulse, which was followed by another 90° pulse after the time TR. The 90° pulse "tilts" the existing longitudinal magnetization into a transversal plane, resulting in a transversal magnetization. The more longitudinal magnetization we have, the stronger the initial transversal magnetization immediately after the 90° pulse. As we read earlier, with a very long TR, all tissues will have recovered their longitudinal magnetization totally; differences in T_1 of the tissues examined will not influence the signal,

as enough time passed by to allow even tissues with a long T_1 to relax totally.

In the spin echo sequence we start with a 90° pulse, which also tilts what is there as longitudinal magnetization (it does not matter, that there were other pulses such as the 180° pulse in between). When we choose a long TR, as we just said, then differences in T_1 do not really matter. When we also use a short TE, differences in signal intensity due to differences in T_2 have not had enough time to become pronounced yet. The signal, that we get, is thus neither T_1 - nor T_2 -weighted, but mainly influenced by differences in proton or spin density. The more protons, the more signal, if you look at it in a simple manner (figure 40).

Fig. 40

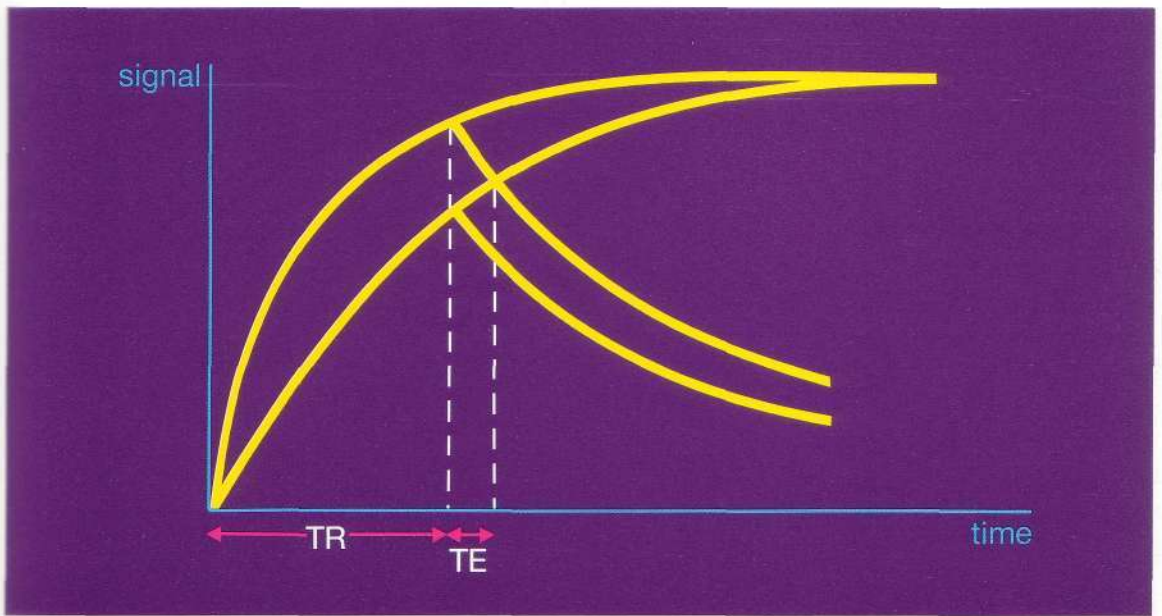
By combining T_1 - and T_2 -curves signal intensity of certain tissues can be determined for a pulse sequence using TR and TE as illustrated, and as explained in figure 39. What happens, when we choose a long TR, as illustrated? With a long TR, differences in T_1 , in longitudinal magnetization time are not very important any more, as all tissues have regained their full longitudinal magnetization. When we only wait a very short TE then differences in signal intensity due to differences in T_2 have not yet had time to become pronounced. The resulting picture is thus neither T_1 - nor T_2 -weighted, but mostly determined by the proton density of the tissues (for this, ideally TE should be zero).



And when we use a long TR and a long TE?

With a long TR there are no prevailing differences in T_1 . With the long TE however, differences in T_2 become pronounced (figure 41). Thus the resulting picture is T_2 -weighted.

Fig. 41
When we wait a long TR and a long TE, differences in T_2 have had time enough to become pronounced, the resulting picture is T_2 -weighted.



What if we use a shorter TR and a short TE?

With a short TR, tissues have not recovered their longitudinal magnetization, thus differences in T_1 (which determines, how fast longitudinal magnetization is regained) will show up in form of signal intensity differences (fig. 42). When TE is short, differences in T_2 cannot really manifest themselves, so the resulting picture is still T_1 -weighted (there is a lower limit for TE, because it takes some time for the 180° pulse to be sent in and do its effects).

What if we use a very short TR and a very long TE?

This is only a theoretical question. Why? With a very short TR, there will be only very little longitudinal magnetization which is "tilted". And with a long TE we even allow the small amount of transversal magnetization resulting to disappear to a large extent. The resulting signal will be so small, of so little intensity, that it cannot be used to make a reasonable picture.

Fig. 42

When we wait a shorter time TR, differences in T_1 influence tissue contrast to a larger extent, the picture is T_1 -weighted, especially when we also wait a short TE (when signal differences due to differing T_2 s have not had time to become pronounced).

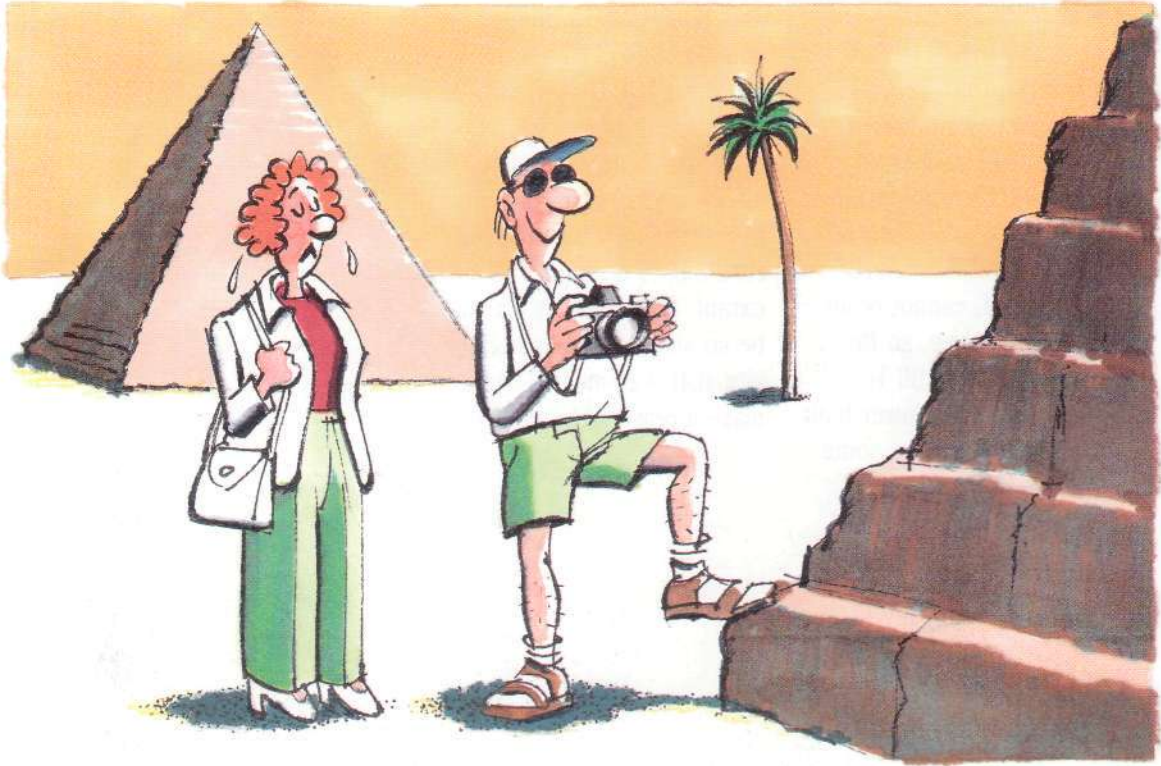
If you have not been concentrating

for the last minutes, you will probably be just short of giving up right now. How to remember this - even if you do not understand all of it (which hopefully is not the case)?

Try looking at fig. 43. What do you see? A man with short TRousers. And considering the weather conditions, this makes only one person in the picture happy.

This should remind you that a short TR (ousers) gives a T_1 -weighted image (only I is happy).

Fig. 43
What to choose for a T_1 -weighted image?



What do you see in figure 44?
The same couple is having tea.
Now having tea, which is usually
served hot, always takes long.
And in the illustration the long
TEa makes two people happy.
This should remind you that a
long TE gives a T_2 -weighted
image.

Fig.44
What to choose for a T_2 -weighted image?

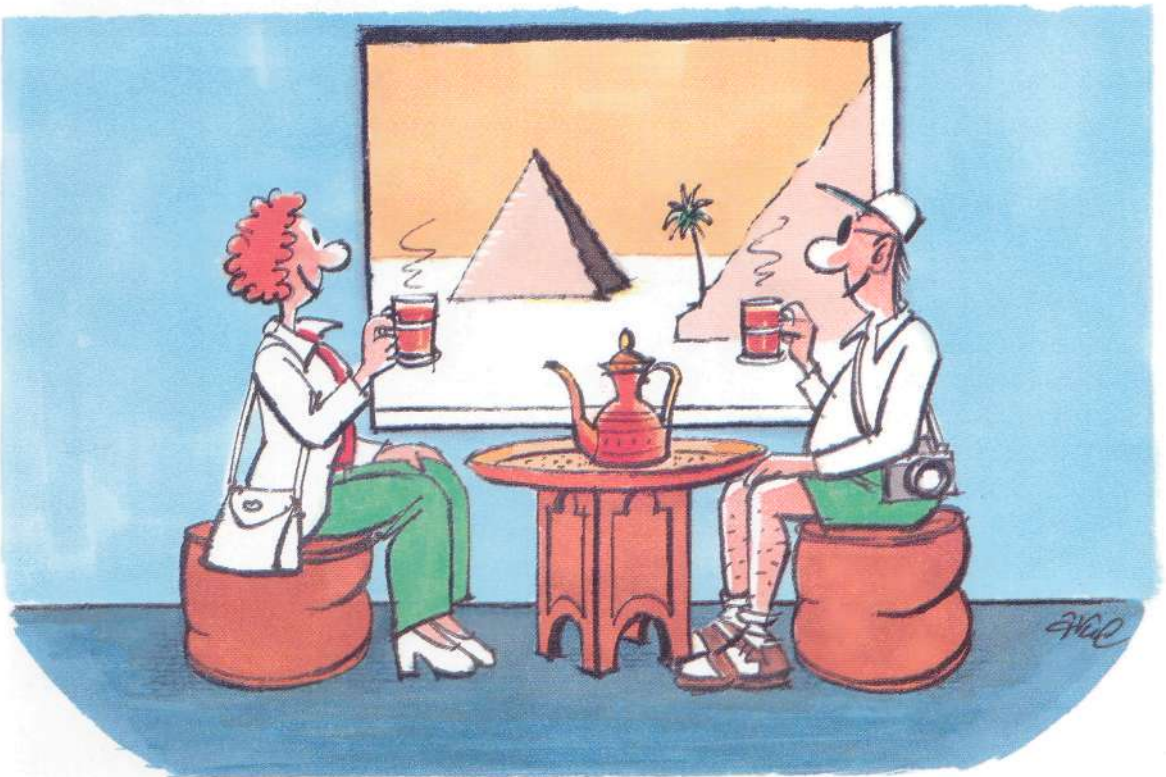
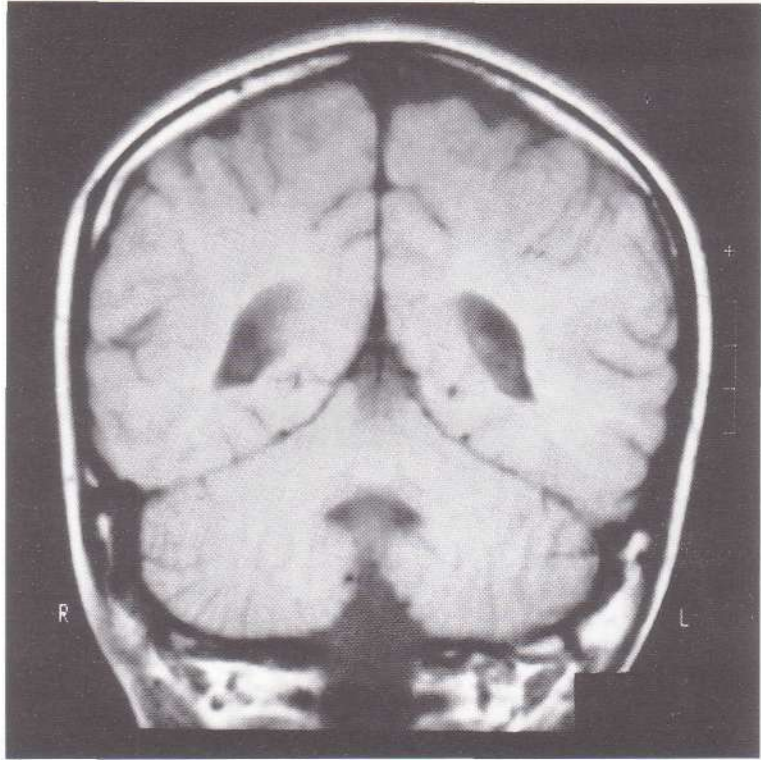
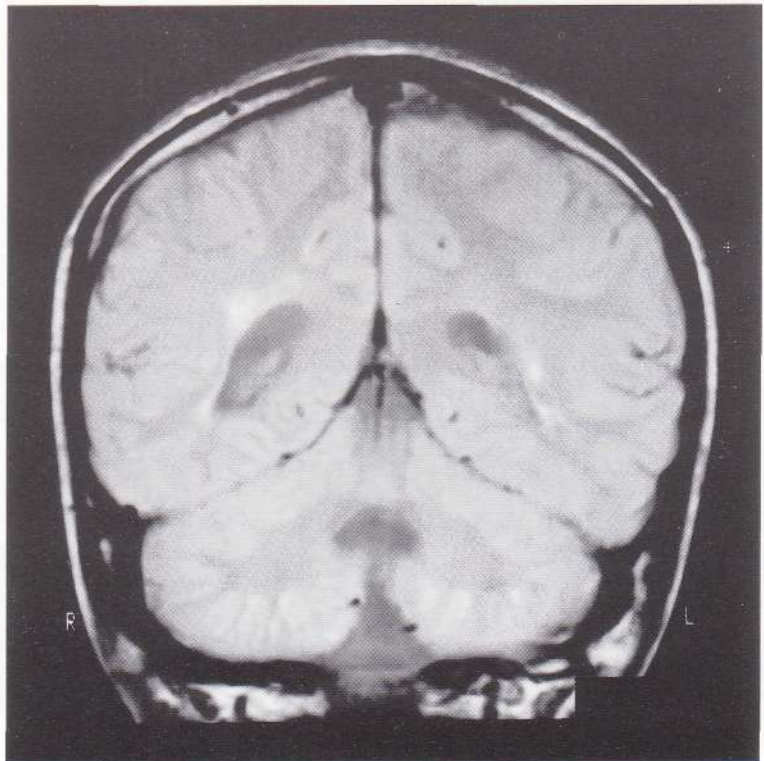


Fig. 45

T₁- (a), proton (spin) density- (b), and T₂-weighted (c) images of the same patient. The CSF is black on the T₁-weighted image. However, it has the strongest signal in the T₂-weighted image. On the spin-density image it is of intermediate signal intensity.

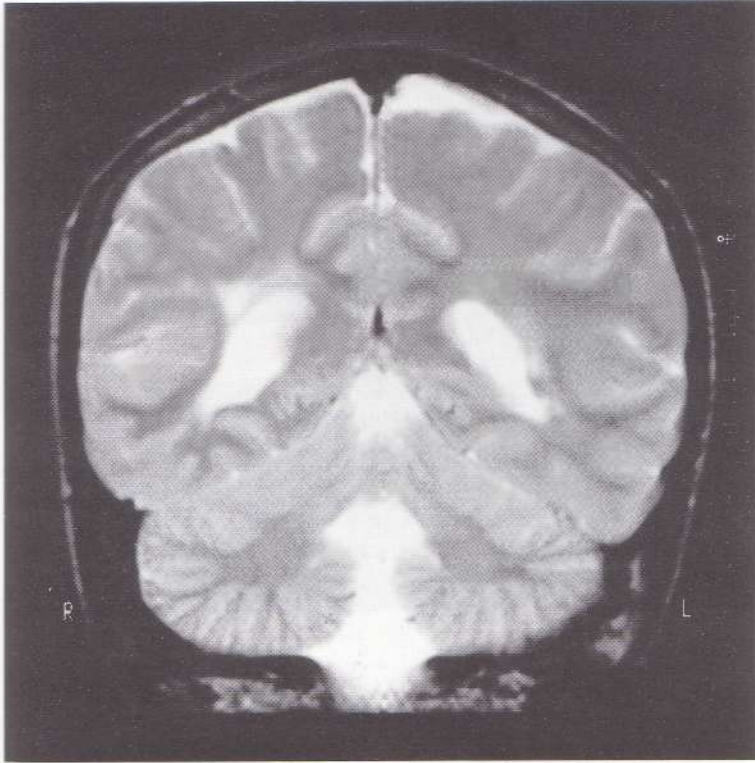


a



b

Some practical hints to image interpretation



C

How can we tell from a picture, whether it is a T_1 - or a T_2 -weighted image, when imaging was done with a normal pulse sequence, not one of the fast sequences (see below). As a rule of thumb: if you see white fluid, e.g. CSF or urine, you are dealing with a T_2 -weighted image. If the fluid is darker than the solids, we have a T_1 - or a proton-density image. Look at fig. 45: in (a) CSF is dark, the grey substance is darker (more grey) than the white substance; this is a typical T_1 -weighted image. In (b) CSF still is dark, even though its signal intensity is slightly higher than in the T_1 -weighted image; contrast between grey and white substance is reversed. This is a proton/spin density-weighted image, and as grey substance has a higher water content, i.e. contains more protons, its signal intensity is higher than that of the white substance. In (c) CSF has a higher signal intensity than grey and white substance, the picture is T_2 -weighted.

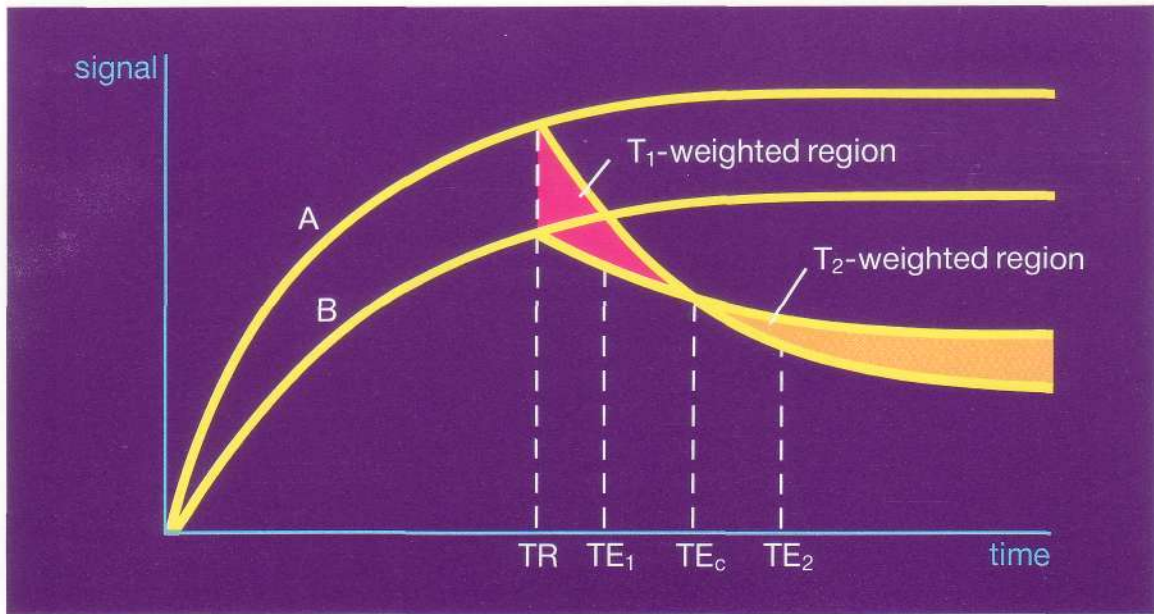


Fig. 46

T_2 -curves of different tissues can intersect. The signal intensity of the tissues is reversed choosing a TE beyond the crossing point (TE_c): before this crossing point (e.g. at TE_1) tissue A has a higher signal intensity than tissue B. This means that image contrast is still determined by differences in T_1 : the tissue A with the shorter T_1 has the stronger signal intensity. At TE_c both tissues have the same signal intensity, and thus cannot be differentiated. After the crossing point (e.g. at TE_2) the relative signal intensities are reversed, and tissue B has the stronger signal.

These are rules-of-thumb only. Actually, to be really sure, you would have to look at two pictures taken with different imaging parameters. Why? Look at figure 46. You see that in this example the T_2 -curves start at different "heights", and cross each other (they do not have to run parallel, it is just better to understand them in the beginning, when they do and for didactic reasons they have been depicted like that). The fact that the curves intersect is very important:

- with a TE before the crossing point (TE_1), tissue A will have a higher signal intensity
- with a TE right at that point, (TE_c), we cannot distinguish the tissues at all, as they have the same signal intensity.

Thus, you might be unlucky, and choose a pulse sequence with just those imaging parameters that do not allow tissue differentiation (which is the reason for performing two different examinations with different T_1 - and T_2 -weightings).

- with a TE beyond the crossing point (TE_2), tissue A will have a lower signal than tissue B.
- before this crossing point (which you do not know, looking at a picture normally), the relative signal intensities are still governed by differences in T_1 . The tissue with the shorter T_1 (or the higher proton density, if we have a long TR) still has the higher signal intensity. Only with longer TEs does the T_2 -weighting come up. Think about that for a moment!

Now we have already heard about many parameters, that influence the MR picture, so T_1 , T_2 , proton density, pulse sequences, TR and TE - but there are more, e.g. flow, and contrast media.

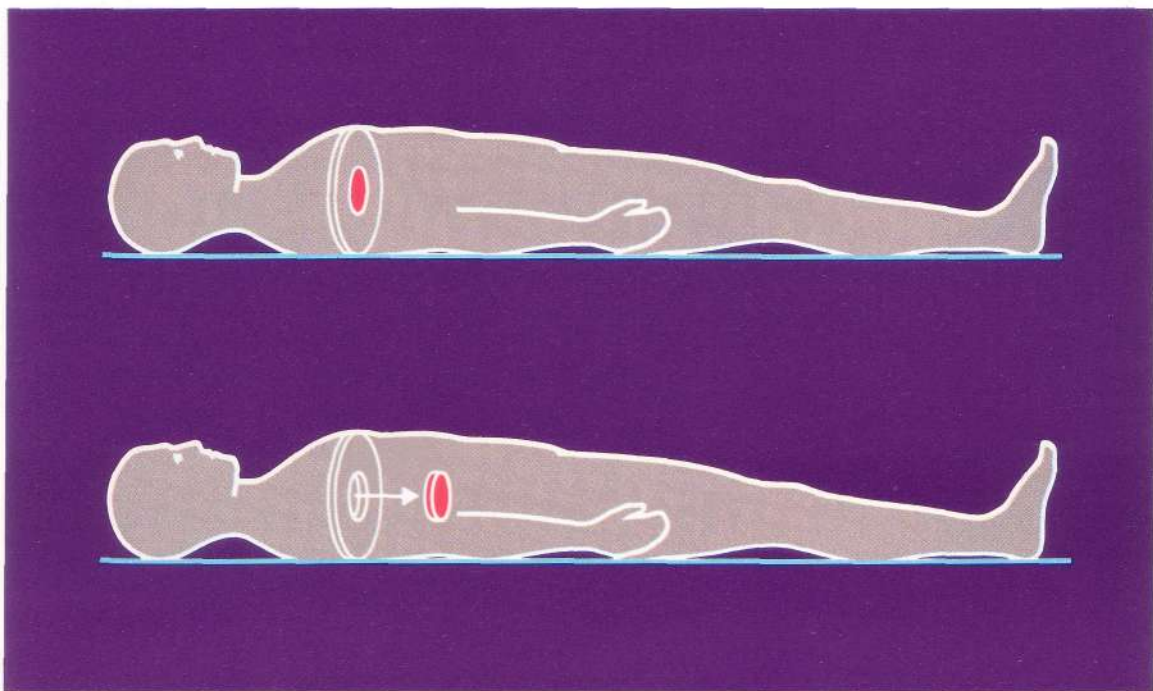
How does flow influence the signal?

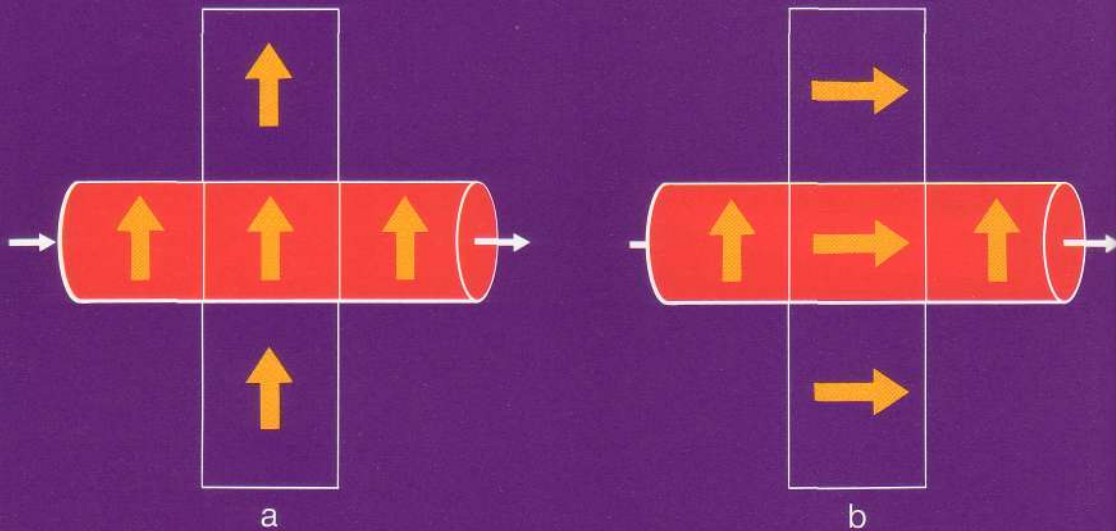
The fact that flow influences the MR signal has been known for a long time. The first experiments on this subject were carried out about 30 years ago. Interestingly enough, this phenomenon was used to measure flow in the fuel pipes of satellite rockets, without having to put any obstruction into the flow lines.

The subject of how flow influences the MR signal is rather complex and difficult, but let us at least get some idea about it. In fig. 47 we have a body sec-

tion through which a vessel is crossing. When we send in our first 90° pulse, all the protons in the cross section are influenced by the radio wave. After we turn the RF pulse off, we "listen" into the section and record a signal. At this time, all the original blood in our vessel may have left the examined slice. So there is no signal coming out of the vessel; it appears black in the picture. This phenomenon is called flow-void phenomenon.

Fig. 47
Flow effects are responsible for the black appearance of flowing blood, the signal void in blood vessels.





This is not the only way in which flow may influence the picture, there may be all kinds of things, e.g. also flow-related enhancement.

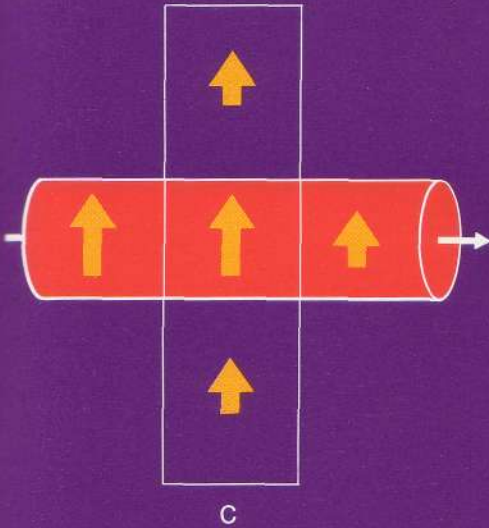
Take a look at figure 48, which shows a blood vessel going through a slice which is being examined, (a) represents the situation before the 90° pulse and (b) immediately after the pulse. If we wait some mope time before we send in a second 90° pulse, like in (c), protons will have undergone some relaxation, and there is some longitudinal magnetization again, as shown by the arrows pointing back up. The protons in the bloodvessel, however, have left the slice and been

replaced by protons that still have all of their longitudinal magnetization. If we send in a second 90° pulse now, there will be more signal coming from the vessel than from its surroundings, because there is more longitudinal magnetization at this time.

The whole subject with signal strength and flow effects is even much more complicated. For example, when you do multislice imaging; i.e. take images of more than one slice at the same time (see page 85), the signal also depends on the direction of the flow. In addition, it differs over the cross section of a vessel, depending on the flow profile, and whether there is laminar or turbulent flow. If you want to know more about this, you should look it up in one of the comprehensive standard text books.

Fig. 48

Flow can have differing effects on signal intensity, and can also cause flow-related enhancement, which is explained in detail in the text.



They will also give you more information on MRI angiography. In this technique the fact that flow influences the MRI signal is used in a beneficial manner by displaying the moving protons.

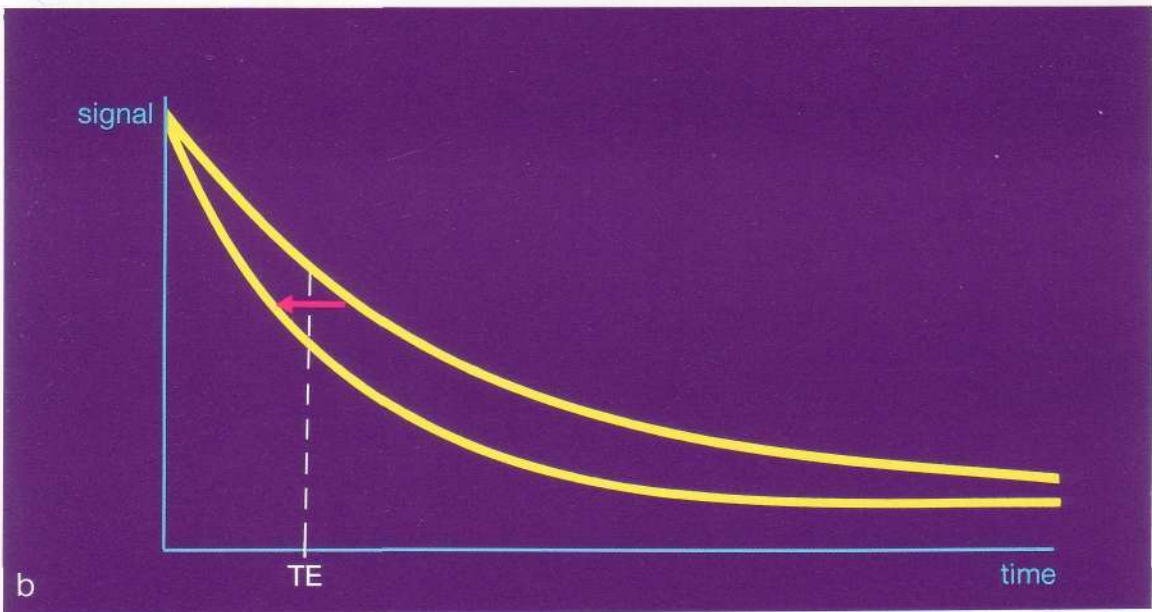
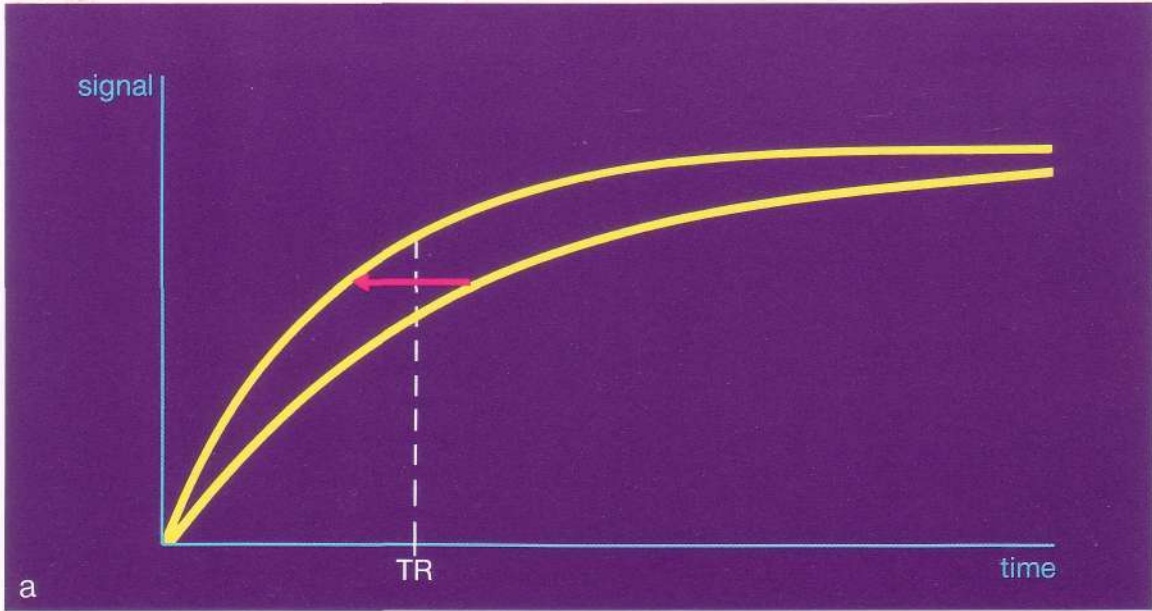


Fig. 49

Paramagnetic substances like Gadolinium shorten the T₁ and the T₂ of their surroundings. The respective T₁- and T₂-curves are shifted towards the left.

In effect, that means that for a certain TR there is more, for a certain TE there is less signal.

Now what about MR contrast media?

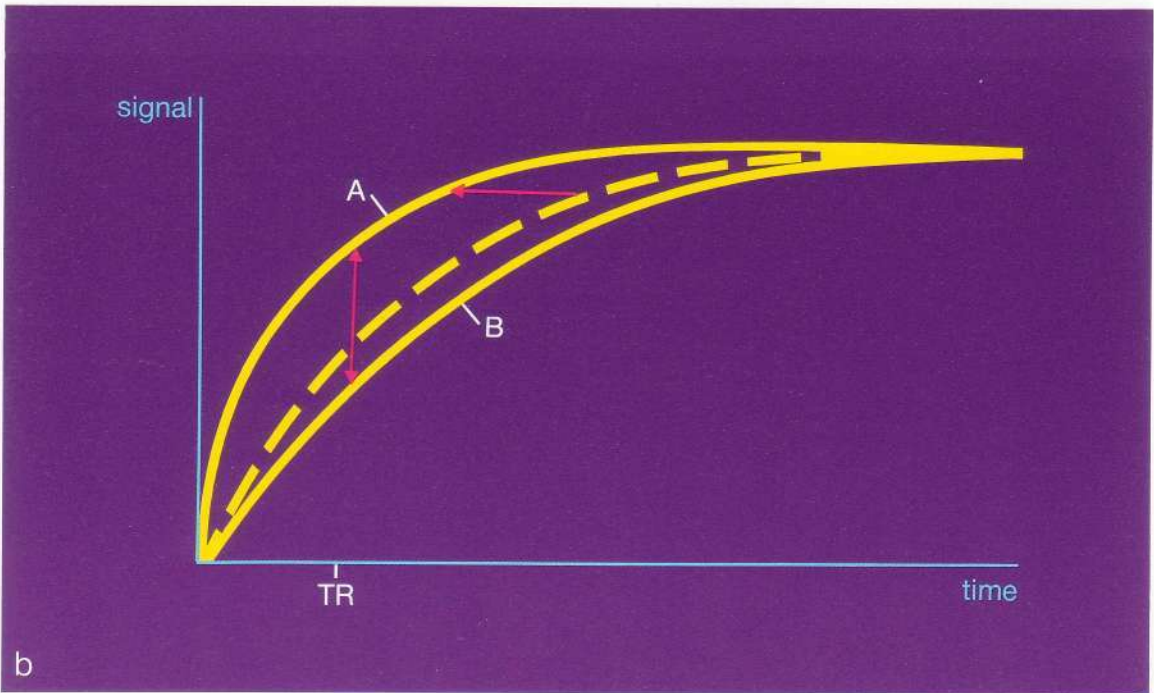
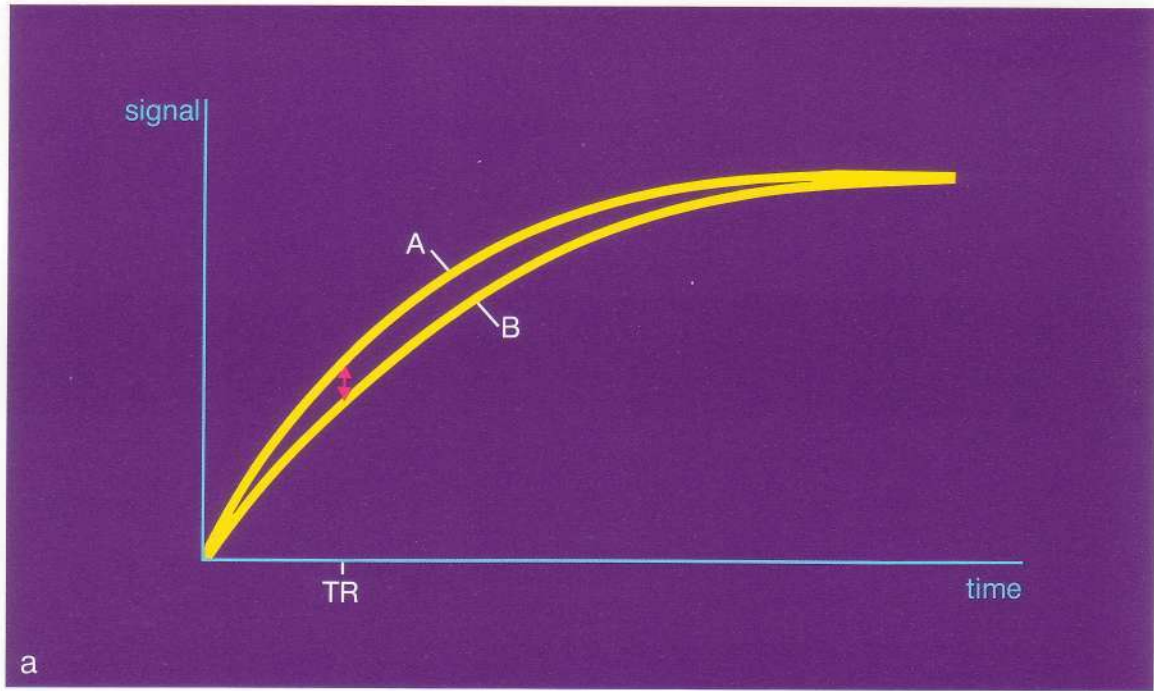
Certain so-called paramagnetic substances have small local magnetic fields which cause a shortening of the relaxation times of the surrounding protons. This effect is named proton relaxation enhancement. The body contains paramagnetic substances under normal circumstances. Examples are degradation products of hemoglobin, e.g. deoxyglobin and methemoglobin, which are found in hematomas, or also molecular oxygen.

Gadolinium, a paramagnetic substance, is used as an MR contrast medium (Magnevist®).

Chemically the substance is a rare earth. As Gadolinium is toxic in its free state, it is bound to DTPA in a certain way (chelation), which solves the problem of toxicity.

The effect of the contrast medium is a change of the signal intensity by shortening T_1 and T_2 in its surroundings (fig. 49).





In fig. 50 this is illustrated for two tissues, A and B. The i.v. administered Gadolinium enters tissue A, the T_1 of tissue A becomes shorter and the T_1 -curve is shifted to the left. The result is that the signal from tissue A at time TR is stronger than it was before and the two tissues can be better differentiated, because there is better contrast.

When we perform a T_2 -weighted examination there is less signal coming from tissue A after contrast medium application, because the contrast medium shortens T_2 , and shifts the T_2 -curve to the left.

As loss of signal often is more difficult to appreciate than a signal enhancement, T_1 -weighted images are the predominant imaging technique used after contrast medium injection.

As the substance is not distributed evenly throughout the body, signals from different tissues will also be influenced differently. Vascularized tumor tissues are enhanced for example. It is also important that the Gadolinium does not go through the intact, but rather the disrupted blood-brain-barrier.

It has been shown that the use of contrast media increases lesion detection and diagnostic accuracy of MRI. It may, for example, help with differentiation between tumor tissue and surrounding edema, which might be otherwise indistinguishable. Gadolinium entering into the tumor tissue shortens the T_1 , thus making the tumor bright in a T_1 -weighted image, while the surrounding edema may not be influenced at all.

As Gadolinium shortens T_1 , we are able to shorten TR in our examination (see page 72). And because imaging time depends on TR, as we will see later, imaging then may take less time.

The pharmacological properties of Gadolinium-DTPA are very similar to the ones of contrast media in conventional radiology; Gadolinium-DTPA however seems to be even better tolerated.

Ready for a repetition?

As we know by now, many parameters, e.g. T_1 , T_2 , proton density, pulse sequence, influence the appearance of tissues in an MR picture.

- With a short TR we get a T_1 -weighted image
- With long TE the image is T_2 -weighted
- Flow effects can be variable, and cover the spectrum from signal loss to signal enhancement
- Paramagnetic substances, e.g. the contrast medium Gadolinium-DTPA, shorten T_1 and T_2 of the surrounding protons. This results in a signal increase in T_1 -weighted images and a signal decrease in T_2 -weighted images
- T_1 -weighted imaging is the preferred technique after contrast medium injection.

By choosing the pulse sequence and imaging parameters, like TR and TE, we can get T_1 -, T_2 - or proton density (spin density)-weighted images, as we have already heard. Many different pulse sequences have been developed and we should be familiar with their basic concepts. So let us take a look at them.

Fig. 50

In (a) the T_1 -curves for tissue A and B are very close to each other, resulting in only a small difference in signal intensity between the tissues at TR. In (b) the T_1 -curve of tissue A is shifted to the left, as contrast agent entered tissue A but not tissue B. At the same time TR there now is a much greater difference in signal intensity, i.e. tissue contrast.

Partial saturation/ Saturation recovery sequence

Pulse sequences that use 90° pulses only, are the saturation recovery pulse sequence and the partial saturation sequence (fig. 51). We have already discussed them, but we did not give them a name. Basically, the sequences are the same: they consist of two 90° pulses. The difference is in the time interval between pulses, the TR (see page 47). Look at figure 52 with the T_1 -curves (going uphill!) of two different tissues. If we send in the second pulse after a long time, TR_{long} , both tissues

have regained longitudinal magnetization. With a TR_{long} , the saturation recovery sequence (the protons have relaxed, are saturated), the signal is influenced by the proton density (Do you recall the stories with the short trousers and the long teas?). With a TR_{short} , the partial saturation (protons have not relaxed) the T_1 becomes important for the signal intensity, so we get T_1 -weighted pictures, (fig. 52)

Fig. 51
Schematic illustration of the partial saturation/saturation recovery sequence.

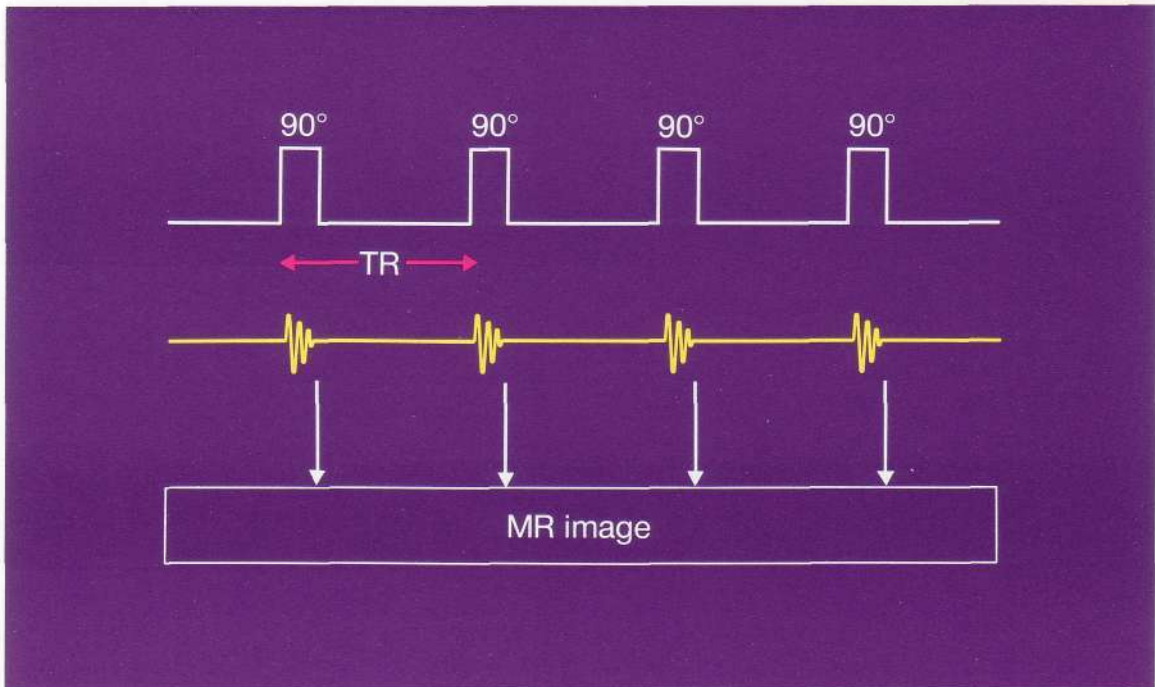
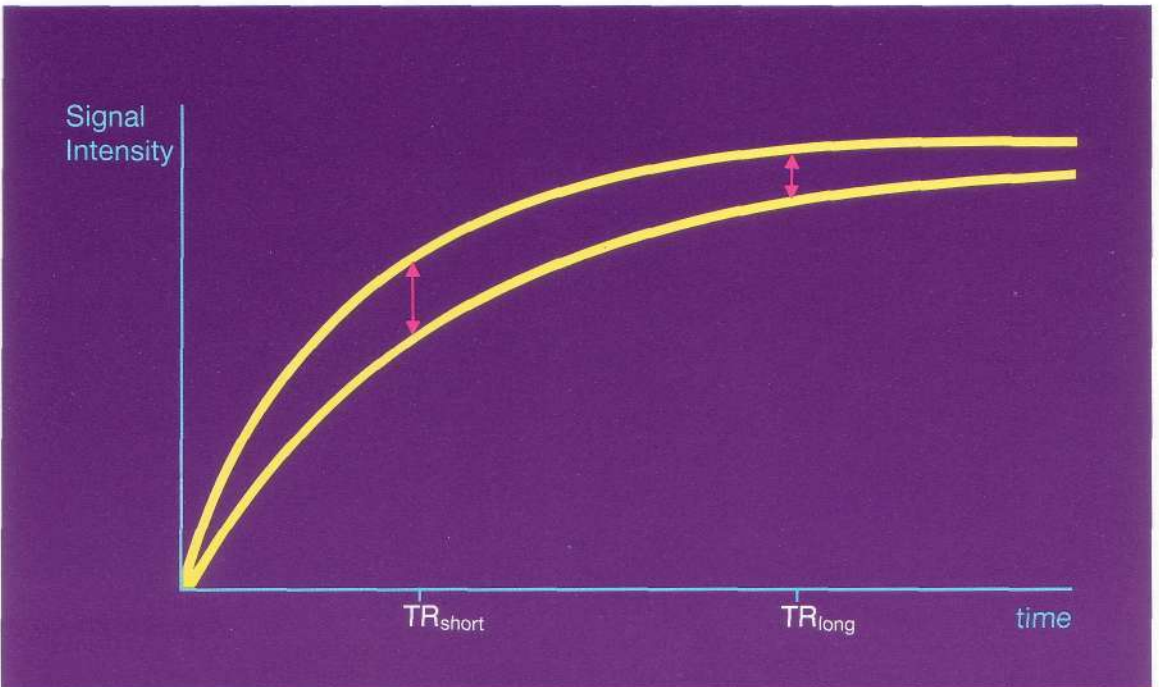


Fig. 52

Signal intensity of tissues having a different T_1 depending on the choice of TR: With a long TR, the saturation recovery sequence, image contrast is determined mainly by proton (spin) density. With a shorter TR, the partial saturation sequence, the resulting image is T_1 -weighted.



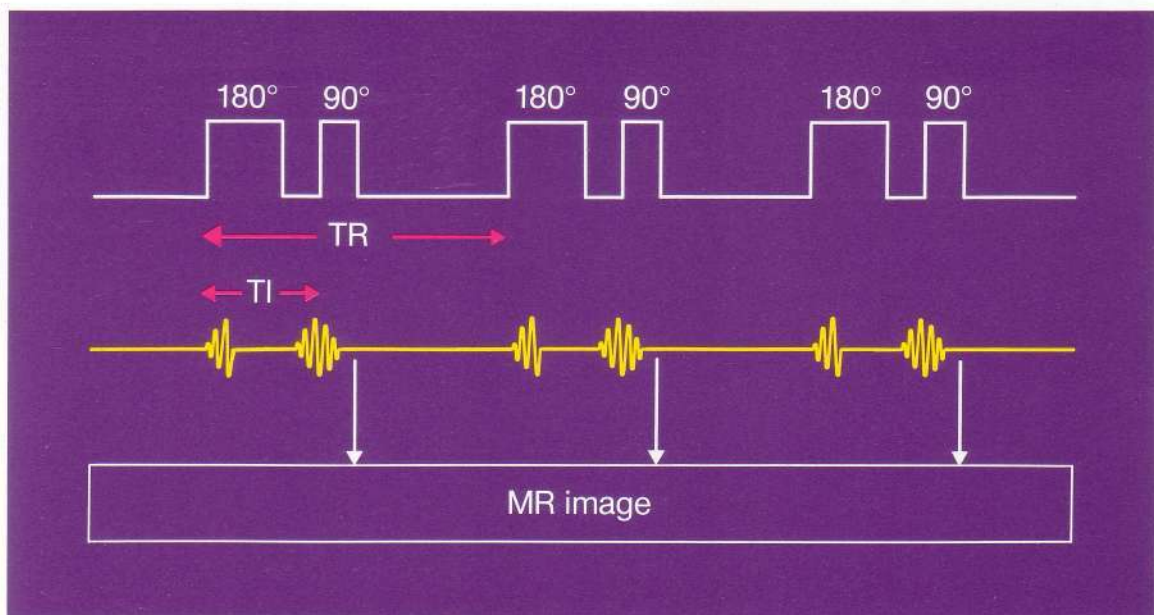


Fig. 53
Schematic illustration of the inversion recovery sequence.

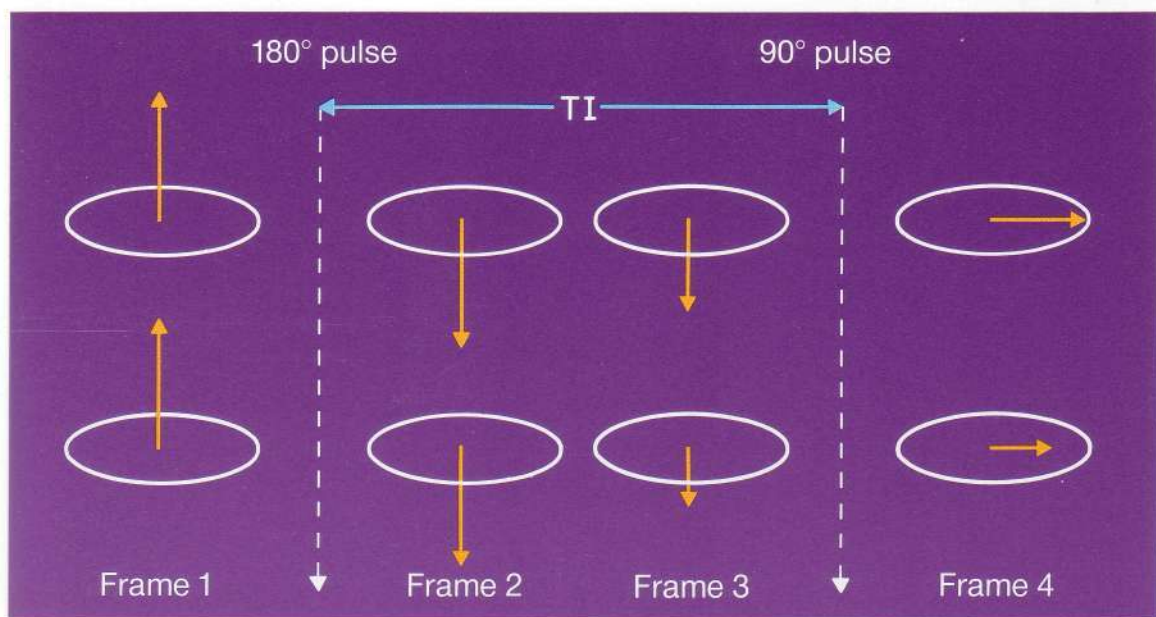


Fig. 54
The inversion recovery sequence uses a 180° pulse which inverts the longitudinal magnetization, followed by a 90° pulse after the time TI. The 90° pulse "tilts"

the magnetization into the transversal (x-y-) plane, so it can be measured/received. The tissue in the bottom row goes back to its original longitudinal

magnetization faster, thus has the shorter T_1 . For the time TI, which is illustrated, this results in less transversal magnetization after the 90° pulse.

Inversion recovery sequence

In contrast to the spin echo sequence, the inversion recovery sequence uses first a 180° pulse which is then followed by a 90° pulse (fig. 53). What happens? The 180° pulse turns the longitudinal magnetization in the opposite direction (all protons that were responsible for the net magnetic moment pointing up, now point down). This is illustrated in fig. 54 for two tissues with a different T_1 (The tissue with the faster longitudinal relaxation, i.e. the shorter T_1 , is in the bottom row). If we do not do anything else, the longitudinal magnetization will slowly go back up, like a ball, that is thrown into

water. To get a measurable signal, however, we need some transversal magnetization. And for this we use the 90° pulse.

The signal that we get depends on the time between the 180° - and the 90° pulse, the time after the inversion by the 180° pulse; this time is thus called

TI = inversion time.

TR is the time between the sequences, as in the other pulse sequences. The signal intensity in an inversion recovery image is dependent on T_1 , which determines how fast the longitudinal magnetization goes back to its original value. So we get a T_1 -weighted image, which is even more T_1 -weighted than partial saturation recovery images.

Spin echo sequence

We have talked about the spin echo sequence in detail already.

It is composed of two pulses: a 90° and a 180° pulse (fig. 55). At this time, you should be able to recall what happens?

The 90° pulse establishes transversal magnetization, however, this is not used to produce an image. Some time ($TE/2$) after the 90° pulse, we sent in a 180° pulse, which rephases the protons that are getting out of phase. After the time TE , we get an echo.

Fig. 55
Schematic illustration of a spin echo pulse sequence. This is repeatedly illustrated as the spin echo sequence is so important.

As we have heard, we can produce not only one, but several echoes. The disadvantage is however, that the signal becomes weaker and weaker.

What were the imaging parameters that influenced the MR signal in the spin echo sequences?

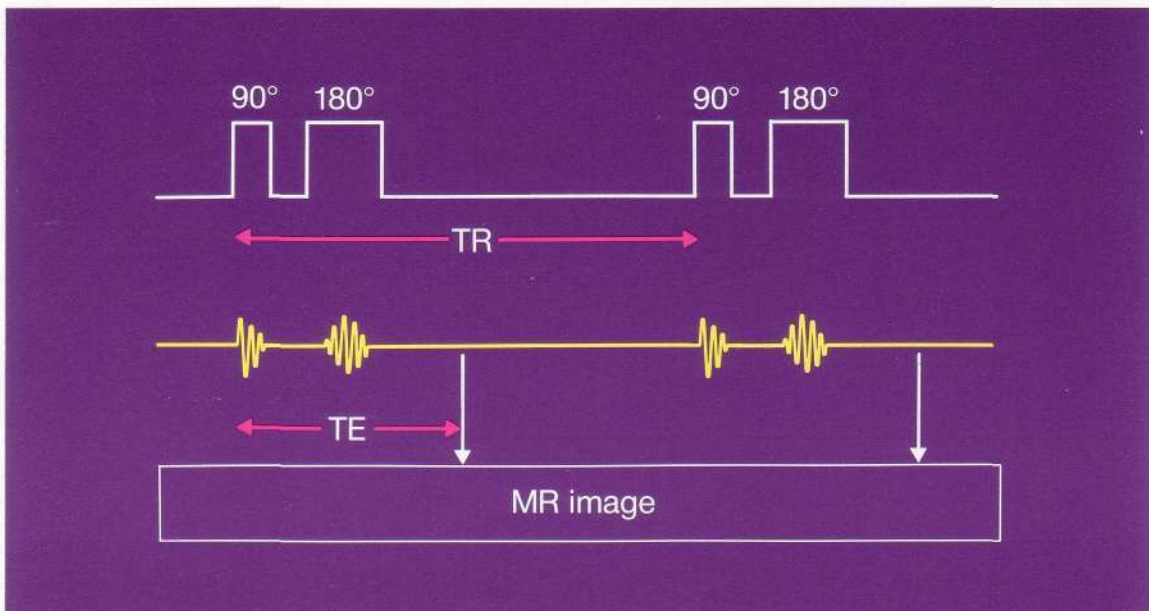
These were:

- TE: the time between the 90° pulse and the echo.
- TR: the time between two pulse sequences, i.e. from one 90° pulse to the next.

What did the TE and the TR do?

They determined how the resulting image was weighted: TE was responsible for the T_2 -weighting, TR for the T_1 -weight-

ing. If you do not remember or even understand this by now, you should read pages 50 to 65 again.



What about those fast imaging sequences?

As the normal imaging sequences take quite some time (the reason is described below), only a limited number of patients can be examined. It is also often very difficult for the patient to lay still for a long time, and image quality decreases with movement. In addition, there is some unavoidable motion, like respiration and heart beat.

To help with these problems, pulse sequences were developed which take less time. Most of these have strange names such as FLASH (Fast Low Angle Shot), or GRASS (Gradient Recalled Acquisition at Steady State). These sequences are becoming more and more important. However, they are much more difficult to understand than the sequences we have talked about up to now. Here is a rough outline.

The TR is the most time consuming parameter of an imaging sequence (see also pages 58 and 85). It makes sense to shorten TR if we want to make imaging faster. And this is done in the fast imaging sequences. But with a decreasing TR there are some problems:

- with a spin echo sequence we used a 180° pulse to refocus the dephasing spins.

Unfortunately, we cannot use a 180° pulse for this purpose when we do imaging with a very short TR: it requires some time to deliver a 180° pulse, and with a very short TR there will not be enough time between the 90° pulses.

- with decreasing TR, longitudinal magnetization will have recovered less and less between pulses (see pages 60-63); so there is only very little longitudinal magnetization to be tilted by the next pulse, yielding very little signal.

These problems are solved as follows:

- we use a different way to refocus the dephasing spins: instead of a 180° pulse, we apply a magnetic field gradient. This means that an uneven magnetic field, a gradient field, is added/ superimposed on the existing magnetic field.

The magnetic field gradient is switched on for a short time. This results in even larger magnetic field inhomogeneities in the examined slice.

(The magnetic field inhomogeneities that exist already at that time are due to inhomogeneities of the external magnetic field, and the internal magnetic field inhomogeneities inside of the tissues, which we talked about earlier - if you do not remember this, go back to page 29 for a short repetition).

Due to these larger magnetic field inhomogeneities, transversal magnetization, and thus the signal, disappears faster (protons dephase faster!).

Then the magnetic gradient is switched off, and after a short time turned back on with the same strength, but in opposite direction. The faster moving protons now become the ones that move slower and vice versa (similar to what happens after a 180° pulse). This results in some rephasing, and thus the signal increases again to a certain maximum, which is called a gradient echo. After this echo the signal decreases again.

What to do about the second problem, the small amount of longitudinal magnetization with a short TR?

The 90° pulse, e.g. in a spin echo sequence, abolishes longitudinal magnetization; longitudinal magnetization however, starts to recover immediately after the 90° pulse, depending on the T_1 of the tissue examined (if you have forgotten, see page 40). The trick with the fast imaging sequences is not to use a 90° pulse, but pulses that cause smaller "flip angles" (mostly in the range of 10° - 35°). With these flip angles smaller than 90 degrees, longitudinal magnetization is not totally abolished. Instead, there is always a substantial amount of longitudinal magnetization left, which can be "tilted" by the next pulse; this gives a reasonable signal even if the next pulse comes in after a very short TR.

As these fast imaging sequences become increasingly important, we should spend a little more time with them. As we have heard (see page 50), a 180° pulse normally "neutralizes" the effects of external magnetic field inhomogeneities. The decay of transversal magnetization is then due to so-called T_2 -effects (see fig. 35).

When we do not use such a 180° pulse, the protons experience larger magnetic field inhomogeneities and get out of phase faster. Signal intensity decays faster, and in this case is due to so-called T_2^* -effects (pronounced: T_2 star-effects), which is also illustrated in figure 35.

Besides these T_2^* -effects, other factors, e.g. the flip angle, influence signal intensity in the fast imaging sequences, which are also called gradient echo sequences for obvious reasons.

Here are some guidelines about gradient echo imaging:

- larger flip angles produce more T_1 -weighting
- longer TEs produce more T_2^* -weighting
- with fast scans there are often intense signals coming out of the vessels.

We save imaging time because

- with small flip angles we only need an RF pulse of short duration
- we do not use a 180° refocussing pulse (which takes time to generate, and exert its effects)
- we do not have to wait long TRs for enough longitudinal magnetization to reappear, as with small flip angles there is always a reasonable amount of longitudinal magnetization left after the initial pulse.

With these fast scans it is possible to do imaging in a second or even less.

Time to repeat and take a break:

- Partial saturation and saturation recovery sequences use 90° pulses. TR is relatively short with partial saturation and relatively long with saturation recovery. While saturation recovery yields proton (spin) density images, the images are T₁-weighted with partial saturation.
- In the inversion recovery sequence a 180° pulse is followed by a 90° pulse, resulting in T₁-weighted images
- A spin echo sequence has a 90° pulse, which is followed by one (or more) 180° pulse(s), to rephase the dephasing protons resulting in one (or more) spin echo(es). This sequence can give proton density-weighted, T₁-weighted, or T₂-weighted images. This is determined by the imaging parameters which are chosen (TR, TE).
- Fast imaging sequences use flip angles that are smaller than 90°, and so-called gradient echoes. Image weighting is also determined by the type of sequence and the imaging parameters chosen.

About imaging time

Is there no other way to decrease imaging time than to use fast sequences?

What actually determines the imaging time?

For MR imaging with normal pulse sequences this can be easily calculated; the acquisition time (a.t.) is:

$$\text{a.t.} = \text{TR} \times \text{N} \times \text{N}_{\text{ex}}$$

This looks a little complicated but is not. Let us start at the back.

N_{ex} is the number of excitations. What does that mean?

For some reasons it is necessary to use not only one signal

measurement, but to repeat the measurement several times. As the MR signal coming out of the patient is very weak, it may be good to add up signals from several measurements, take several "averages", to get a good quality image. Actually, what you get is an image with a better signal-to-noise ratio. Naturally, imaging time increases with every additional measurement.

To illustrate this:
Just imagine that you are sitting in a large audience, where many people make noise. Someone sitting next to you whispers something in your ear, but you cannot really understand him, because there is so much background noise. What you will probably do, is ask him to repeat what he said one or several times. You mentally add up the information which you are receiving each time. As this signal is always the same, it will increase by adding it up. The background noise, however, is not always the same. Instead it is random and fluctuates and does not add up the way the

signal does. So all together you will have a better signal-to-noise ratio (which you would also have if the person spoke louder).

What is "N"? As you know from other imaging methods (or your TV), pictures are made of picture elements, which altogether make up the image matrix, e.g. a 256 x 256 matrix has 256 rows of 256 picture elements (pixels). In our equation, N is the number of rows in a matrix, like rows in a letter. The more rows you have, the more time it takes for the image. Just think about this as writing a letter:
if you have paper with 5 rows on a page, you will finish a page

faster than if you have 25 rows to write. However, you have more contents, more detail on a page/picture, when you work with more rows.

Repeated measurements result in a better signal-to-noise ratio.



And why does TR influence acquisition time?

If you choose a long time TR to repeat your pulse sequence, to perform additional signal measurements, imaging takes longer than with a short TR. However, there is a trick that can shorten imaging time.

While we are waiting to repeat our imaging sequence in one slice, i.e. while we wait for TR to go by, (slice A in fig. 56), we might as well make measurements in one or more different slices (slices B, C and D in fig. 56). The longer the TR, the more slices we can excite in the meantime. So for just adding a little extra time, we will examine many slices instead of one, and imaging time per slice decreases substantially.

We perform so called multi-slice imaging.

Another way to possibly reduce TR, and thus imaging time, is the use of a contrast medium: as we have read, Gadolinium shortens T_1 . And when T_1 is shorter, the TR can also be shorter, without a loss in signal intensity of the tissue in question (see fig. 49).

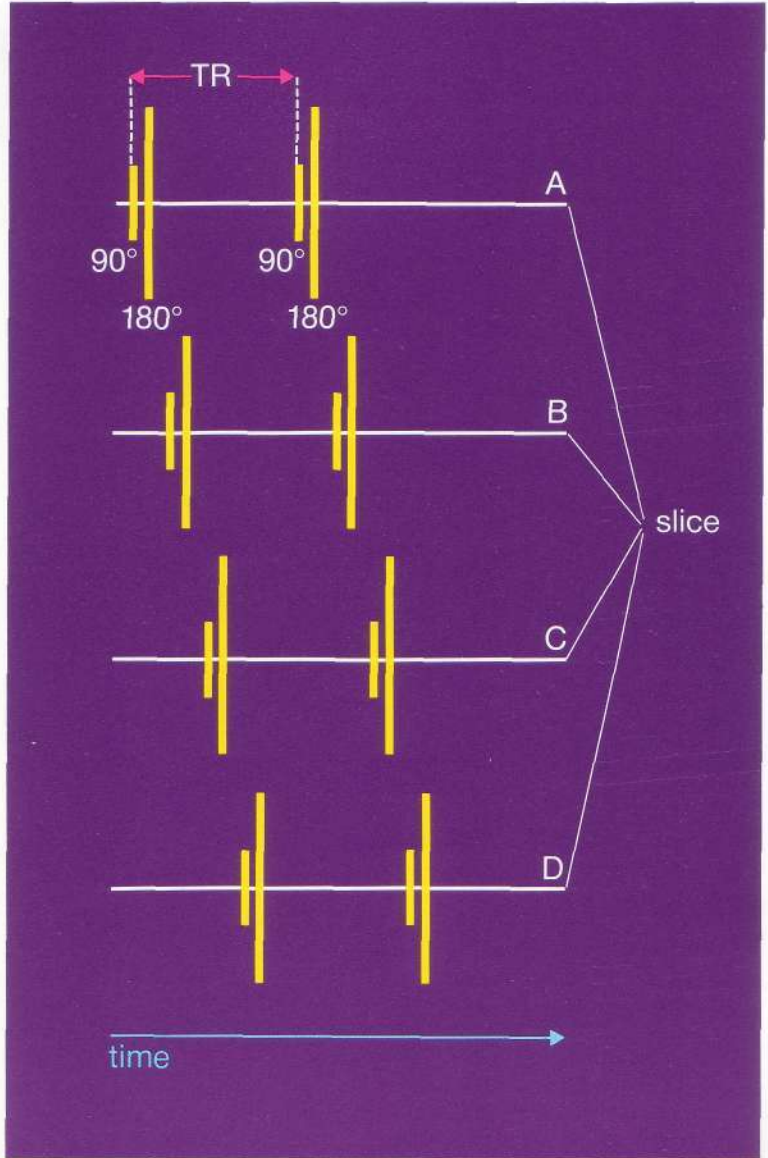


Fig. 56

Multislice imaging: while we wait for the time TR to pass by for another signal measurement in slice A, we perform signal measurements in additional slices (B-D). So during the time TR, we actually recorded signals for more than one image, though from different slices.

Let us review all the factors that influence signal intensity in MR

These are:

- hydrogen density (page 47)
- T_1 (page 26)
- T_2 (page 30)
- flow (page 69)
- the pulse sequence (pages 76-82)
- TR (page 47)
- TE (page 56)
- TI (page 79)
- flip angle (page 82)
- use of contrast medium (page 73)

If you are not sure about one of these, go back to the page cited, and read the text once more. In case you feel familiar with these facts, go on, and read about some important things in MR imaging, that we have not talked about yet.

How can we select a slice which we want to examine?

When we put a patient into an MR scanner he/she is in a rather homogeneous magnetic field. So all the protons in the whole body have the same Larmor frequency, and will be excited/disturbed by the same RF pulse. To examine a specific slice only, a second magnetic field is superimposed on the external field which has different strengths in varying locations. The magnetic field is therefore stronger or weaker in some places than in others (fig. 57). This additional field is called a gradient field, and is produced by the so-called gradient coils. This gradient field modifies the strength of the original magnetic field. In figure 57, magnetic field strength increases for different cross sections from the feet towards the head. Consequently, the protons in the different slices experience different magnetic fields, and thus have different precession frequencies. So the RF pulses which disturb the protons in the different slices must have different frequencies as well.

As gradient fields can be superimposed in any direction, it is possible to define not only transversal slices, but all kinds of different imaging planes without moving the patient. The gradient field that enables us to examine a specific slice is also called slice selecting gradient.

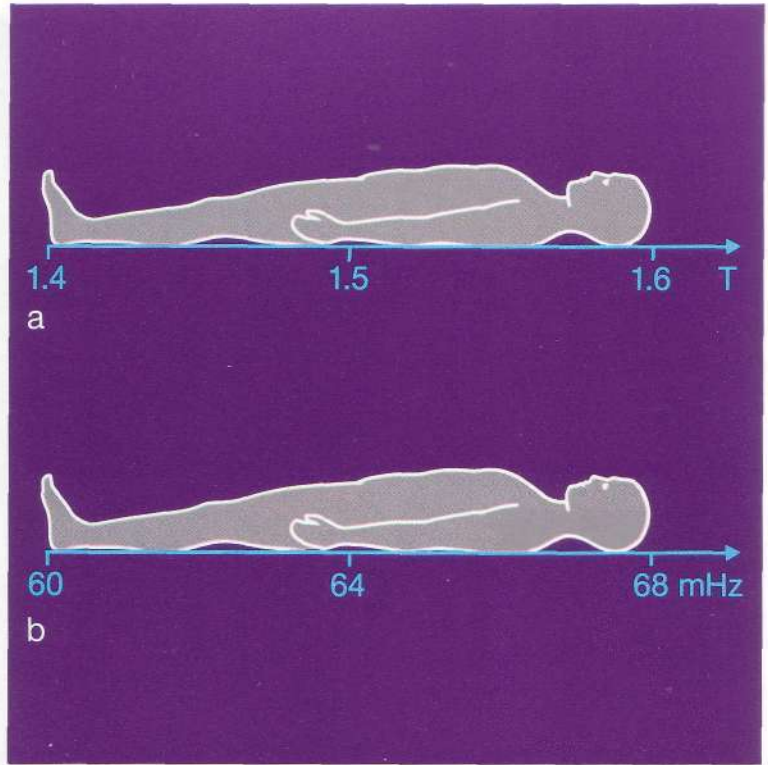


Fig. 57
Magnetic gradient fields are superimposed on the field of the MR magnet, so that different cross sections of the body experience magnetic fields of differing strength. In the illustration the resulting magnetic field strength is increasing from 14 Tesla at the feet, to 16 Tesla at the head. As magnetic field strength and precessing/resonant frequency are directly correlated (Larmor equation), the resonant frequency at the feet is about 60 mHz, while it is about 68 mHz at the top of the head in our example. By selecting a certain RF pulse frequency we determine the location of the slice which we examine.

How can we determine or select a certain slice thickness?

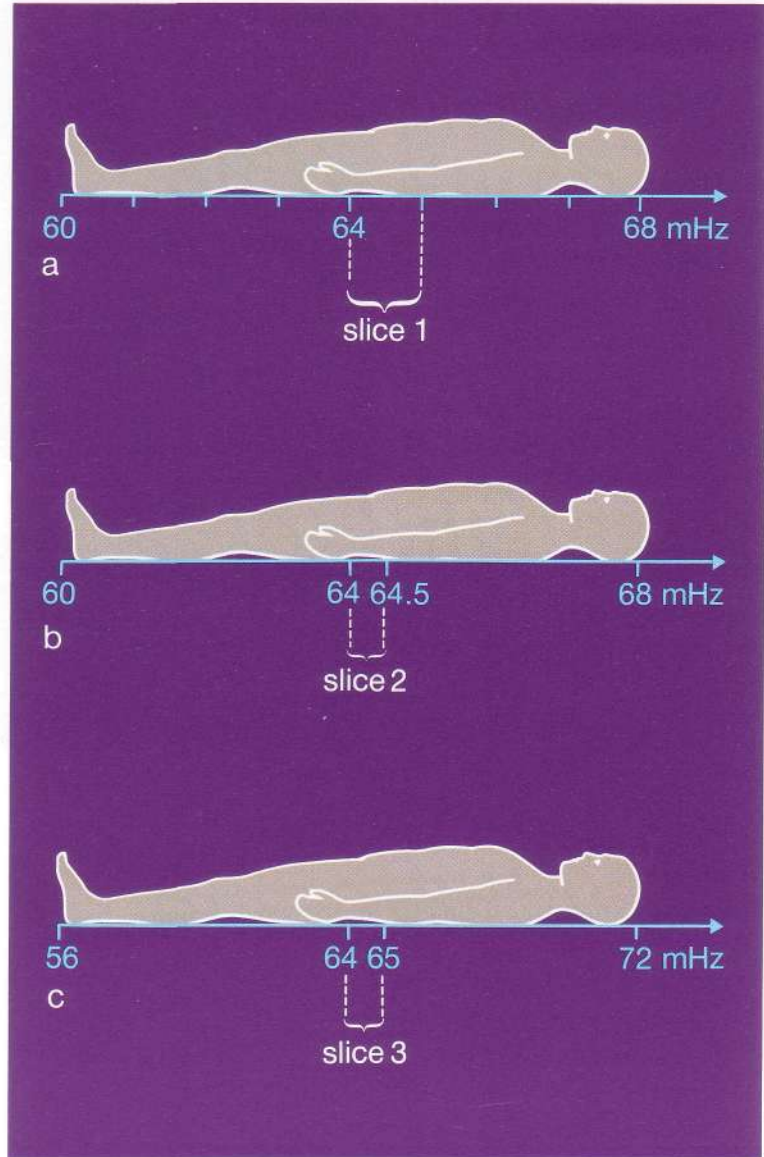
We can select a different slice thickness in two ways (fig. 58):

- we send in not only one specific frequency (which is not done in practice) but an RF pulse that has a range of frequencies; the wider the range of frequencies, the thicker the slice in which protons will be excited. This has been illustrated in figure 58.

If we use an RF pulse with frequencies from 64 to 65 mHz, we will get a slice thickness S_1 (fig. 58a). If, however, we only use frequencies from 64 to 64.5 mHz, the protons in a smaller slice, S_2 , will show resonance (fig. 58b).

- if we use the same range of radio frequencies, the same band width as it is called, slice thickness can be modified by the slope of the gradient field, as is illustrated in figure 58c.

If we have a steeper gradient field, i.e. one that has more difference in field strength over a specific distance, the precession frequencies will also vary to a larger degree.



In figure 58a and c an RF pulse of the same band-width, containing frequencies between 64 and 65 mHz, is used both times. The slice thickness in 58c with the steeper gradient field is, however, smaller than in a.

Fig. 58

There are two ways to determine slice thickness. The first is to use an RF pulse that has not only one specific frequency but a certain range of frequencies, a so-called bandwidth. If, for example, we send in an RF pulse, which contains frequencies between 64 and 65 mHz, protons in slice 1 will be influenced by the RF pulse.

When the RF pulse only contains frequencies between 64 mHz and 64.5 mHz, thus has a smaller bandwidth, slice 2, which is half as thick as slice 1 will be imaged. When there is more difference in magnetic field strength between the level of the feet and the head, i.e. the magnetic gradient is steeper, the resulting slice will be thinner, even though the RF pulse bandwidth is the same. This is illustrated in (c) where the magnetic field strength varies more between the feet and the head than in (a); the corresponding resonant frequencies are 56 to 72 mHz in (c) vs. 60 to 68 mHz in (a). Using the same RF pulse containing frequencies from 64 to 65 mHz results in imaging of a thinner slice 3 in (c) than in (a).

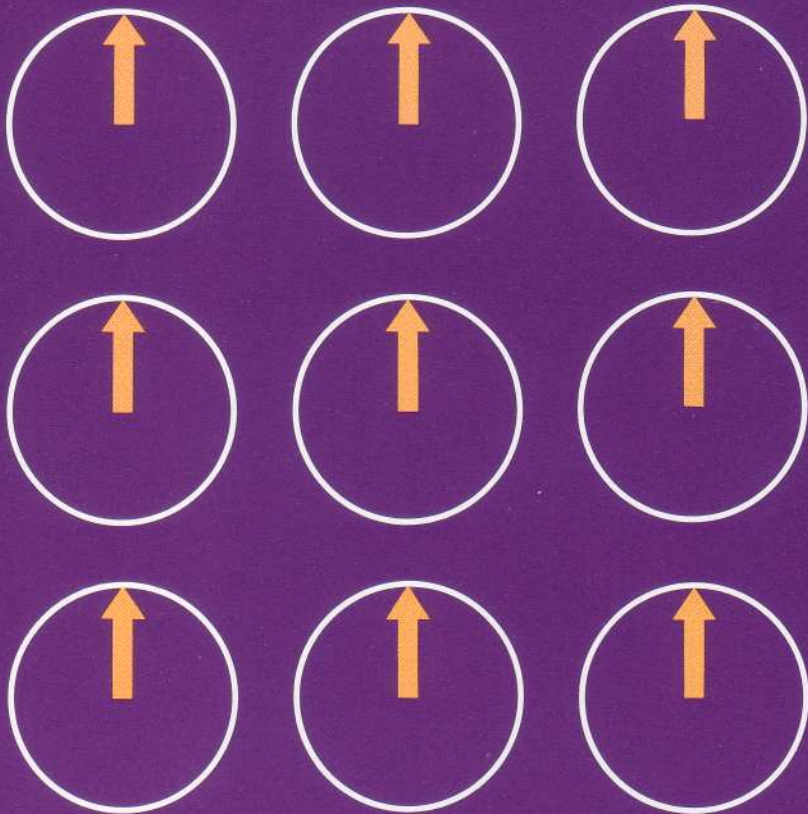
Where does the signal come from?

Now we have selected position and thickness of our slice. But how can we find out, from what point of our slice a certain signal is coming from - some information that we must have to construct a picture?

The trick is similar to the slice selecting gradient which is turned on only during application of the RF pulse.

After the RF pulse is sent in, we apply another gradient field. This is illustrated in figure 59, which shows the situation of the protons in the slice selected, precessing all with the same frequency. We now apply another gradient field which in our example decreases from left to right. So the precession frequency of the protons will also decrease from left to right (in our example the precession frequencies are 65, 64 and 63 mHz, respectively).

The result is that the protons in the different columns emit their signals with these different frequencies. The gradient applied is thus also called the frequency encoding gradient. However, all protons in one column will still have signals with the same frequency. As this is not enough spatial information, we have to do something else. Theoretically, we could use the same trick with the gradients again. This, however, causes some practical difficulties (e.g. this may result in two points at different locations having the same frequency). The problem is solved in a different way this time.



a

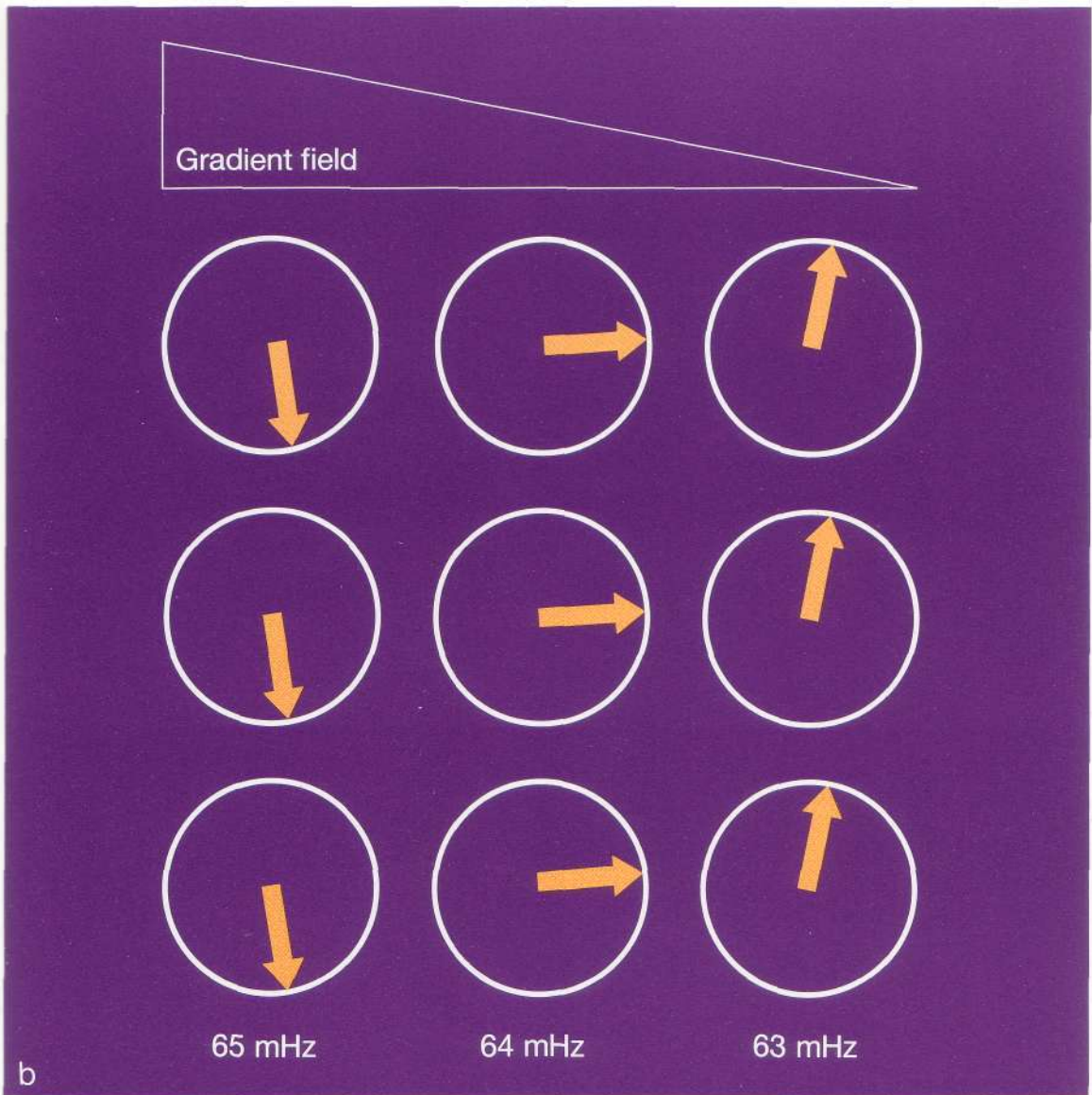


Fig. 59

To determine where in a certain slice a signal comes from we use a magnetic gradient field. In (a) nine protons in the same slice are depicted. They precess in phase with the same frequency after the RF pulse is sent in.

A magnetic gradient field is then superimposed on the external field, which in (b) decreases in strength from left

to right. The protons in the three rows now experience different magnetic fields, and thus give off their signals with different frequencies (e.g. 65, 64, and 63 mHz). The corresponding magnetic gradient is called the frequency encoding gradient. We now can tell from which row a signal comes from, but still cannot pinpoint the exact place of origin.

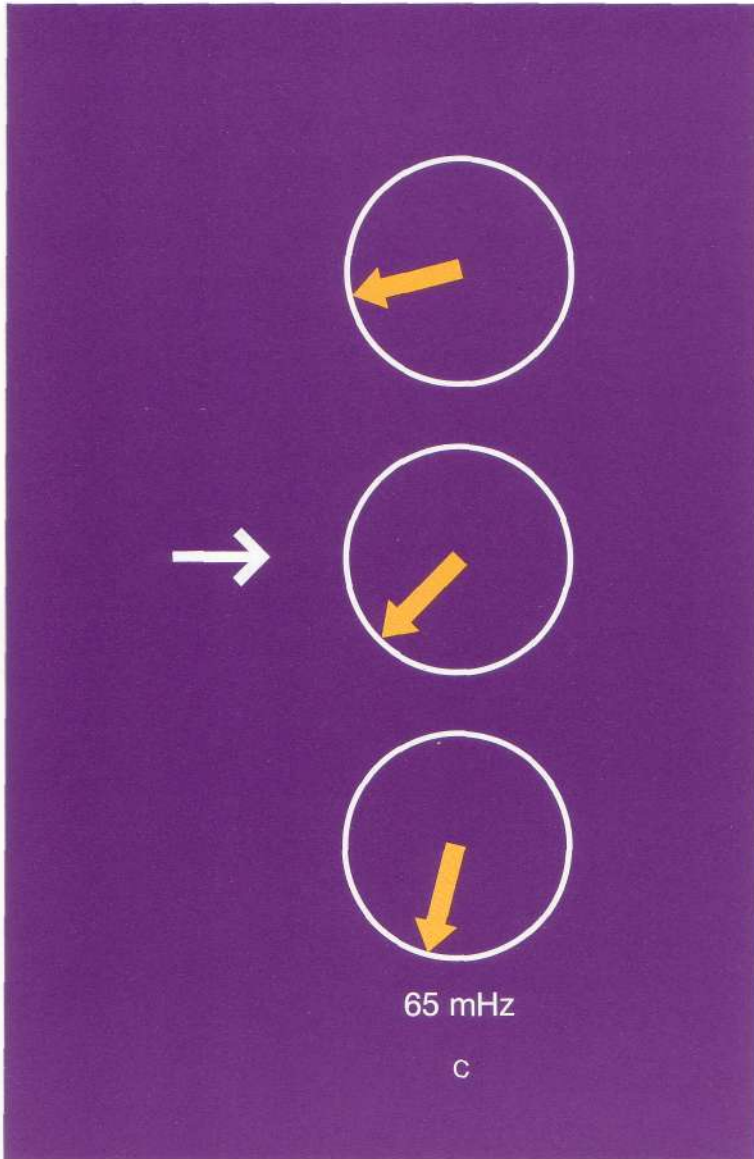


Fig. 60

To find out from where in a row with the same frequency a certain signal comes from, we use an additional gradient. In (a) the row with the precession frequency of 65 mHz from figure 61 is depicted. We now switch on a gradient field, which is stronger at the top than at the bottom of the row for a very short time. The proton at the top thus precesses faster than the one in the middle, which in turn precesses faster than the proton at the bottom. This difference in precessing frequency only lasts for a very short time. However, when the gradient is switched off, all protons experience the same magnetic field again, thus have the same 65 mHz precession frequency again (c). However, now we have a little difference among these protons: even though they precess with the same frequency again, they are a little out of phase, and consequently give off signals of the same frequency, which, however, are different in phase, and because of this can be differentiated. The corresponding gradient is called the phase encoding gradient.

As the gradient which we used causes protons to precess in different phases, it is called the phase encoding gradient.

What finally comes out after we have applied all these gradients is a mixture of different signals. These have different frequencies, and signals with the same frequency have different

phases, all according to their location. By means of a mathematical process called Fourier transformation, a computer can analyze how much signal of a specific frequency and phase is coming out. As these signals can be assigned to a certain location in the slice, we now can reconstruct our image.

Let us repeat:

- we can select a slice to be examined by using a gradient field, which is superimposed on the external magnetic field. Protons along this gradient field are exposed to different magnetic field strengths, and thus have different precession frequencies. As they have different precession frequencies, we can send in an RF pulse that contains only those frequencies, which excite the protons in the slice which we want to image.
- slice thickness can be altered in two ways: by changing the band width of the RF pulse, or by modifying the steepness of the gradient field.

D the slice selecting gradient is only turned on during the RF pulse

- to determine the point in a slice from which a certain signal is coming, we use two other gradients, the frequency encoding and the phase encoding gradient

- the frequency encoding gradient is sent in after the slice selection gradient. It is applied in the direction of the y-axis. This results in different precession frequencies along the y-axis, and thus different frequencies of the corresponding signals

- the phase encoding gradient is turned on for a short time after the RF pulse along the x-axis. During this short time, the protons along the x-axis precess with different frequencies. When this gradient is switched off, they go back to their former precession frequency, which was the same for all of them. Due to this phase encoding gradient, however, the protons and their signals are now out of phase, which can be detected.

(Note: which gradient is applied in what direction (y-axis, x-axis) can be varied!).

- by means of the Fourier transformation, a computer can analyze the mixture of signals that come out of a slice, and determine the intensity of the components that either have different frequencies or different phase

- it is known where in a given slice a signal with a certain frequency or certain phase comes from. And as the Fourier transformation gave us the corresponding signal intensities, we can now assign a certain signal intensity to a specific location, which results in our MR picture (. . . finally!)

A few more basics

By now we have discussed just about every important aspect of MR basics. But: why did we always talk about the protons only? What about the nuclei? As you recall, atoms have a nucleus made up by protons and neutrons. An exception is the hydrogen nucleus, which only consists of one proton. And when we talk about the proton, we talk about the hydrogen nucleus, as both are the same (the terms proton and hydrogen nucleus can thus be used interchangeably). The hydrogen nucleus is best for MR imaging as hydrogen occurs in large abundance throughout the body. Hydrogen also gives the best signal among the nuclei: from an equal number of different nuclei in the same magnetic field, hydrogen gives the most intense signal. All of the routine MR imaging is proton/hydrogen imaging nowadays. However, lots of research is being done on the use of other nuclei.

Can we use every other nucleus for imaging?

The answer is no. We can only use nuclei that have

- a spin, and
- an odd number of protons (and neutrons, but this will go into too much physics, so we will only talk about the protons).

This can be easily explained: as we read in the beginning (page 6), the protons were spinning around, and thus their electrical charge was also spinning, moving. And the moving electrical charge was the current that caused the magnetic field of the proton, which was the basis for everything. If not for the spin, there would be no magnetic field. How to explain the second requirement, the odd number? Just think about the proton as a little bar magnet. If you have a nucleus with two (or any other even number) protons, these little bar magnets would cling together like any other magnets (opposite poles attract).

The result: their magnetic moments would cancel each other out. If we have a nucleus with an odd number of protons, e.g. three, pairs of protons will still cling together and neutralize each other. However, there will always be one proton left that still has a magnetic moment. Nuclei with odd numbers of protons thus have a magnetic moment, and can principally be used for MRI. Examples are ¹³C, ¹⁹F, ²³Na, ³¹P.

Let us have a look at some hardware

The most important part of the MR machine is the main magnet, which has to be pretty strong to allow MR imaging. The strength of a magnet is given in Tesla or Gauss, where

1 Tesla = 10 000 Gauss.

Gauss was a German mathematician, who was the first to measure the geomagnetic field of the earth.

Tesla is considered to be the "father" of the alternating current. He was a peculiar fellow, having refused to share the Nobel prize with the inventor Thomas Edison in the early 1900s.

Magnets used for imaging mostly have field strengths somewhere between .5 to 15 Tesla (as a comparison: the earth's magnetic field is between 0.3 and 0.7 G, the magnet of a refrigerator door has about 100 G = 0.01 T). Their magnetic field has to be very homogeneous, as it directly determines the precession frequency. The homogeneity is quoted in terms as ppm, part per million, in a defined volume (to calculate this, the difference between maximum and minimum field strength is divided by the average field strength and this multiplied by one million). How detrimental even rather small inhomogeneities and thus differences in precession frequency can be, was illustrated on page 29 already. Homogeneity of the magnetic field can be improved by making some electrical or mechanical adjustments, a process called shimming. In MRI different types of magnets are used.

Permanent magnets:

Everybody is probably familiar with a permanent magnet. It is that type of magnet that fascinates little kids. This kind of magnet is always magnetic and does not use any energy for work, which are its advantages. Possible disadvantages are thermal instability, its limited field strength, and its weight (a magnet of 0.3 T may weight about 100 tons!)

Resistive magnets:

In a resistive magnet, an electrical current is passed through a loop of wire and generates a magnetic field. Resistive magnets are therefore also called electromagnets. They are only magnetic as long as there is an electrical current flowing through them. Thus, they use electrical energy. As there is a resistance to the flow of the electricity through the wire, these magnets get warm when in operation, and have to be cooled.

Compared with permanent magnets, they achieve a higher field strength. Resistive magnets are not very practical with very high field strengths because they create lots of heat that must be dissipated.

The relatively new iron core (hybrid resistive magnets) have features of permanent and "normal" resistive magnets, combining some of their advantages.

Superconducting magnets:

Superconducting magnets are the ones most widely used in MR machines at the present time. They also make use of electricity, but they have a special current carrying conductor. This is cooled down to superconducting temperature (about 4° K or -269° C). At this temperature, the current conducting material loses its resistance for electricity. So if you send in an electrical current once, it flows in there permanently, creating a constant magnetic field. So called cryogens (helium, nitrogen) are used for cooling of these magnets, and have to be refilled once in a while.

When for some reason the temperature rises above the superconducting temperature in these magnets, there will be a loss of superconductivity (so-called quench), and sudden resistance to the flow of electricity. This results in rapid heat production, which causes cryogens to boil off rapidly (these leave the system via the so-called quench lines).

Advantages of superconducting magnets are high magnetic field strength and excellent magnetic field homogeneity. (This is in the order of 10-50 ppm over a region 45 cm in diameter). Disadvantages of the superconducting magnets are high costs, and use of rather expensive cryogens.

What is all this talk about the magnetic field strength?
Which is the ideal field strength?

This question is as easy to answer as the question about the ideal horsepower for a car. Here are some of the pros and cons:

- high field strength systems have a better spatial resolution and may be used for spectroscopy
- low field systems on the other hand offer better tissue contrast, are cheaper in price and in operating costs.

Another piece of the hardware: the coils

In MRI radio frequency coils are necessary to send in the RF pulse to excite the protons, and to receive the resulting signal. The same or different coils can be used for transmission of the RF pulse and receiving the signal. A variety of coils are in use.

Volume coils

Volume coils are used in all MR units. These completely surround the part of the body that is to be imaged. These volume coils should be close to the size of the subject. The body coil is a permanent part of the scanner, and surrounds the patient. It is important, as it is the transmitter for all types of examinations. It also receives the signal when larger parts of the body are imaged. The helmet-type head coil acts as receiver coil, the body coil transmitting the RF pulses.

Shim coils

As we have already mentioned in connection with the magnets, magnetic fields have inhomogeneities. Better homogeneity can be achieved by electrical and mechanical adjustments. For this process, which is called shimming, the shim coils are used.

Gradient coils

Gradient coils are used to systematically vary the magnetic field by producing additional linear electromagnetic fields, thus making slice selection and spatial information possible (see pages 86 to 93). As we have three dimensions in space, there are three sets of gradient coils. As these coils bang against their anchoring devices, they are the cause of noise that you can hear during a MR examination.

Surface coils

Surface coils are placed directly on the area of interest, and have different shapes corresponding to the part to be examined. They are receiver coils only, most of the received signal coming from tissues near by; deeper structures cannot be examined with these coils. As with the head coils, the RF pulse is transmitted by the body coil in these cases.

Why do MR units require special facilities?

The large static magnetic field of an MRI system limits system location, as it extends outside of the imager. It can attract metallic objects, and influence mechanical and electrical devices, like computers, monitors, pacemakers and X-ray units. On the other hand there are also external influences. The whole air is full of radio waves - just think about all the stations which you can receive on your radio. To prevent interferences between outside radio waves and those from the MR unit, the whole system is shielded by a Faraday cage. Besides external RF generators, larger metallic objects, especially when moving (elevators, cars), deserve to be mentioned as they can influence the magnetic field.

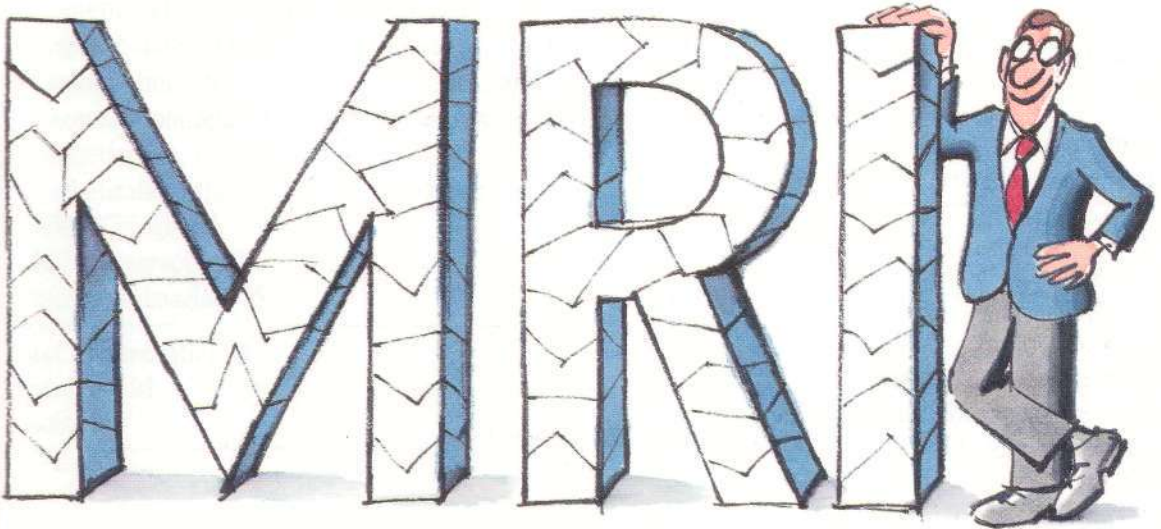
A final look at spectroscopy

MR spectroscopy has been in use for a long time, long before MR was used for imaging. The procedure is used as an analytical tool, as it can identify various chemical states of certain elements without destruction of the sample. It is hoped that spectroscopy and imaging may be combined in the future. This would enable us to obtain in vivo information about the chemistry and metabolism in specific locations. As these measurements could be repeated without harm, follow up studies of cell physiology would be possible. This, for example, may be useful in the evaluation of many diseases and the effects of therapy. As spectroscopy requires magnets with higher field strengths, it can only potentially be performed with the use of MR units which have superconducting magnets. Other magnets cannot do imaging as well as spectroscopy. At the present time there are many people who believe that the spectroscopic potential of MR imaging is even more important than its potential for anatomical imaging.

The final review

Now that you have made it up to here, it is our sincere hope that you know a little bit (more?) about MRI. A final review?

Yes, but let's try a somewhat different approach this time. Take a look at the index on the following pages. Check and see if you understand all of the terms mentioned. If not, refer back to the page numbers listed for a short review.



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References and suggested readings

1. Cohen, M.D.: Pediatric Magnetic Resonance Imaging. W. B. Saunders Company, Philadelphia - London - Toronto, 1986
2. Friedmann, B.R., Jones, J.P., Chavez-Munoz, G., Salmon, A. P., Merritt, C.R.: Principles of MRI. McGraw-Hill, New York - St. Louis - San Francisco, 1989
3. Horowitz, A. L.: MRI Physics for Physicians. Springer Verlag, Berlin - Heidelberg - New York, 1989
4. Kean, D., Smith, M.: Magnetic Resonance Imaging. Williams & Wilkins, Baltimore, 1986
5. Lufkin, R.B.: The MRI Manual. Year Book Medical Publishers. Chicago - London, 1990
6. Partain, C.L., James, A.E., Rollo, F.D., Price, R.R.: Nuclear Magnetic Resonance Imaging. W. B. Saunders Company, Philadelphia - London - Toronto, 1983
7. Runge, V. M.: Enhanced Magnetic Resonance Imaging. C.V. Mosby Company, St. Louis - Washington - Toronto, 1989
8. Sigal, R.: Magnetic Resonance Imaging. Springer Verlag, Berlin - Heidelberg - New York, 1988
9. Stark, D.D., Bradley, W.G.: Magnetic Resonance Imaging. C.V. Mosby Company, 1988
10. Young, S.W.: Nuclear Magnetic Resonance Imaging. Raven Press, New York, 1984