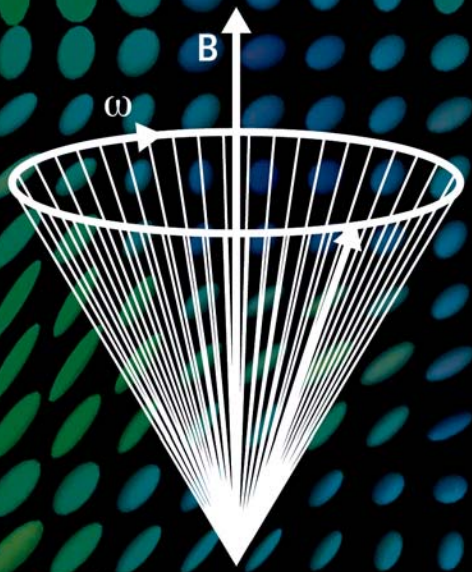
SIEMENS



Magnets, Spins, and Resonances

An introduction to the basics of Magnetic Resonance

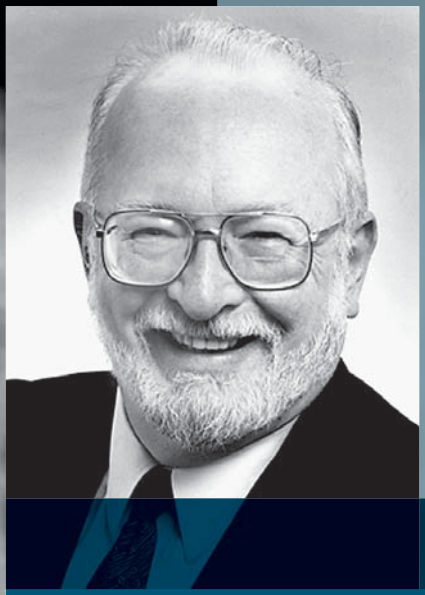
Magnets, Spins, and Resonances

Magnets, Spins, and Resonances

An introduction to the basics of Magnetic Resonance

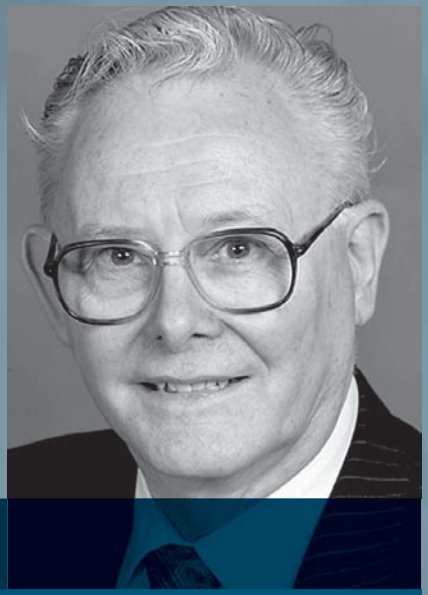
MRI diffusion tensor tractography of the human brain[1] (2015) vs. bell pepper image (1980)

[1] MPI, Leiptzig, Germany



"Before every big breakthrough, it is first a crazy idea."

Paul C. Lauterbur,
Nobel laureate 2003



"It's a different type of image that carries with it much more information about the disease process."

Sir Peter Mansfield,
Nobel laureate 2003

MR Basics: A Long Road Made Easy

For a Nobel laureate, it is a long road to Stockholm. Magnetic Resonance has traveled a long road from probing water droplets and paraffin blocks to imaging bell peppers and the human body, creating seven Nobel laureates along the way. This brochure may shorten your road to understanding the basic principles of MRI.

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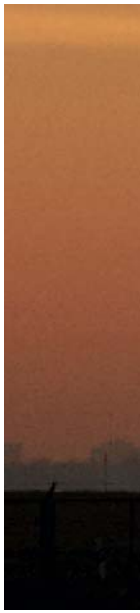


MAGNETOM World

6 Increasing Diagnostic Value

Tracing the MR Signal

How does magnetic resonance imaging work? Let us follow the basic principles step by step. The MR image is created by magnetic reactions in the patient's body that generate a measurable signal.

MRI in a nutshell

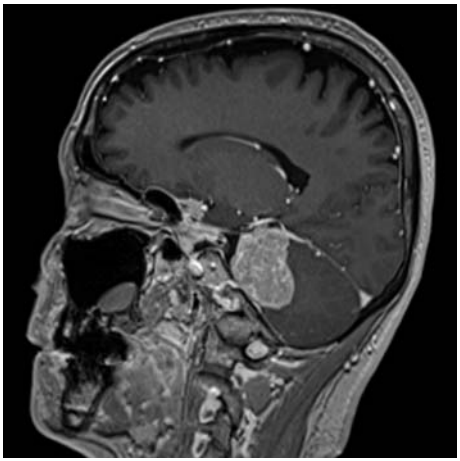
Let us start with a clinical question

How can we detect a lesion in the image?

Medical imaging depends on the ability to visualize anatomical structures, exploiting some physical properties within the human body. For diagnostic purposes, it is essential to discriminate between normal and pathologic tissue.

MR is a non-invasive imaging technique that produces series of slice images with arbitrary angulation, displaying the structure and/or function of the head, body, or extremities. An MRI system may also be used for imaging during interventional procedures.

How and why does MRI work?

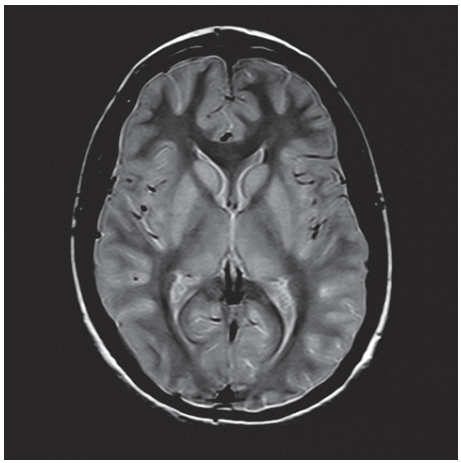


It is all about contrast

Of course, good contrast resolution of the MR image is of paramount importance for a precise diagnosis.

MR imaging offers excellent soft-tissue contrast that results from the combination of several parameters. What is the source in the human body that provides for this contrast?

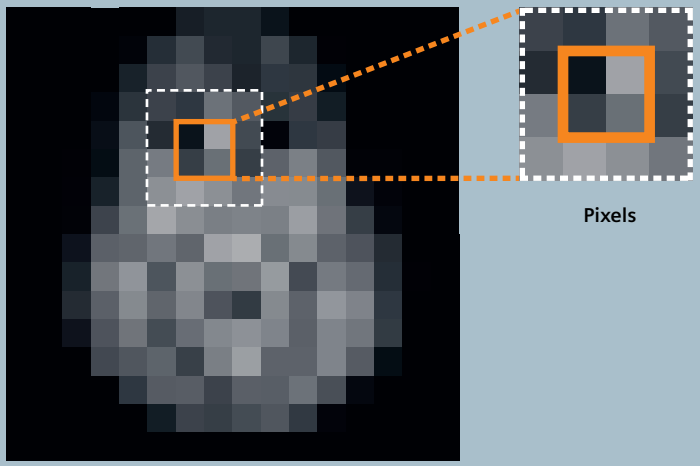
Let us start with an axial slice through a humans head and trace back to where image contrast originates.



"Magnetic resonance measurements may be used as a method for discriminating between malignant tumors and normal tissue."
(Raymond Damadian, 1971)

Tracing image contrast back to the source

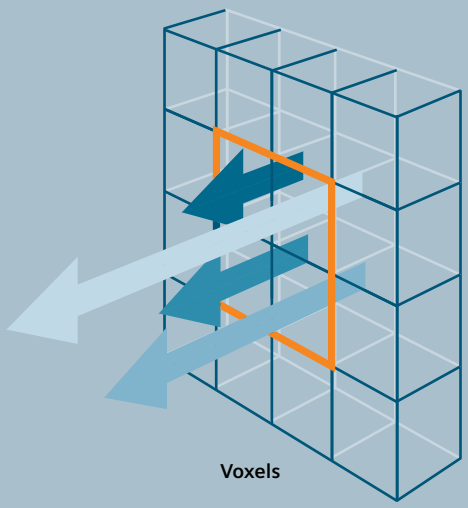
A simplified picture of the MRI “value chain”



The slice image is an array of picture elements (*pixels*). Each pixel has a certain gray value. More pixels in an image mean better resolution.

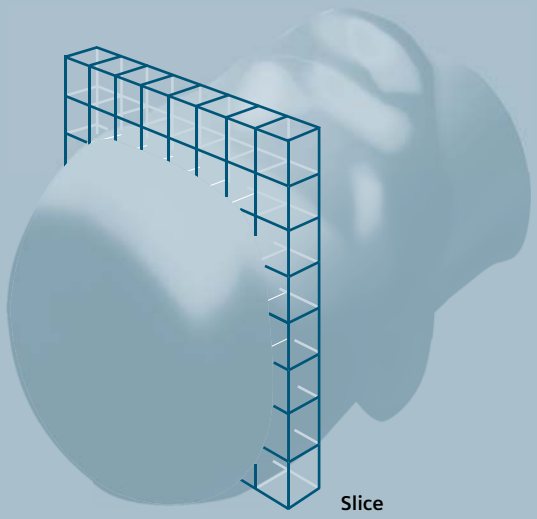
To simplify things, we will start with very low resolution and focus on four contrasting pixels in the brain image (graphic left).

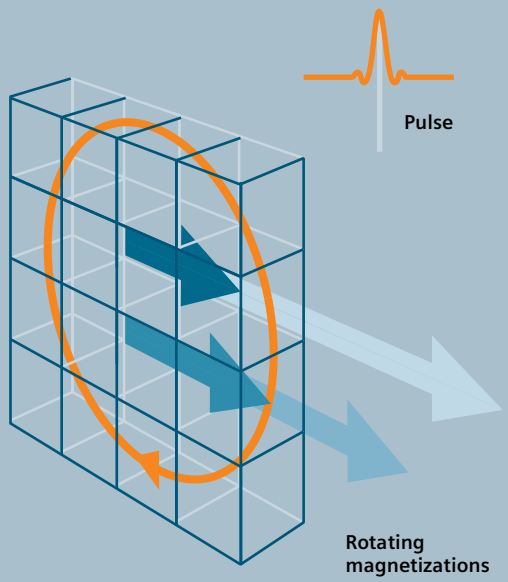
This is crucial: The pixels in the image correspond to volume elements (*voxels*) in the *slice* measured (graphic right).



The patient is positioned in a strong magnetic field. A distribution of "magnetic forces" will build up in the body (arrows left), visualized for only four voxels of interest.

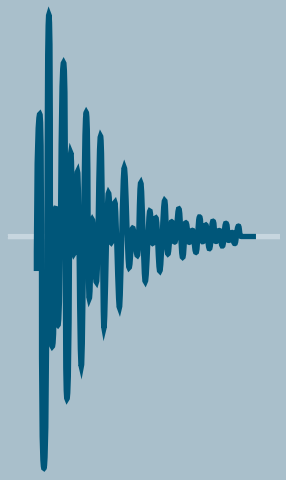
This distribution of magnetism, known as *magnetization*, is not uniform but depends on tissue properties. Stronger magnetization of a voxel is depicted by a longer arrow (left).



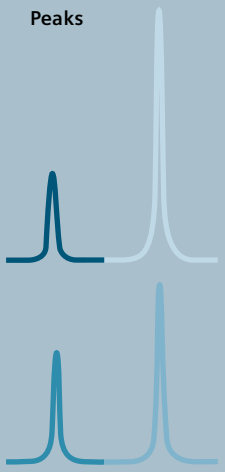


In order to make the magnetic distribution within the body visible, we have to tilt the magnetizations with the help of an electromagnetic *pulse* (left). This behavior is the hallmark of the magnetic resonance process.

The magnetizations begin to *rotate* and generate an alternating electric current in a receiver coil. This current is the *MR signal* (right), a mixture of the magnetizations of the contributing voxels.



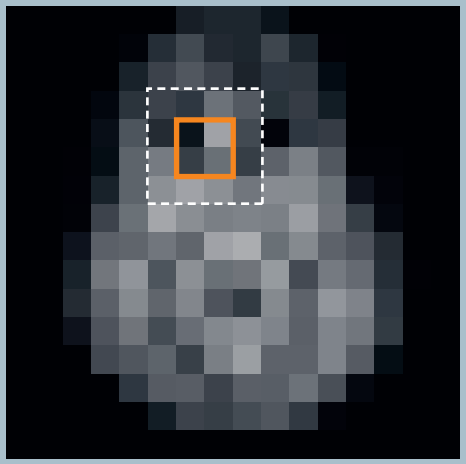
In a magnetic field, the human body builds up a spatially distributed magnetization. Once tilted, the magnetization rotates and generates an MR signal. From a series of signals, an MR image can be computed.



Peaks

Actually, the local magnetizations are made to rotate with different speeds (frequencies) and timing (phases). Hence, we can filter out their individual contributions from the MR signal and thus obtain separated *peaks* (left).

The information of original voxel locations is conserved by the MR imaging method. Thus, we can match individual peaks to corresponding pixels in the image. Peak intensity is assigned to a gray value (right).



Image

Magnetizing the body

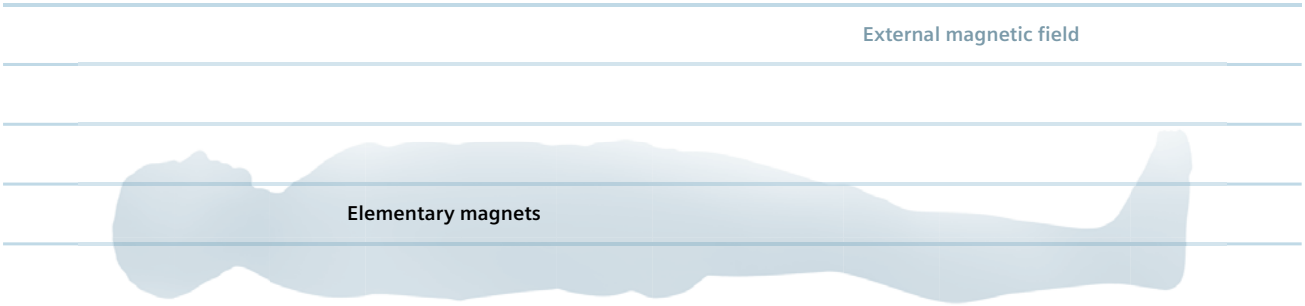
Preparing the patient's body for imaging

Susceptible to magnetic fields

A prerequisite for MR imaging is that the patient is positioned in the strong magnetic field of a scanner. Why?

The human body is *susceptible* to magnetic fields: the strong external field *enhances* the magnetism already present in the tissue.

Let us trace the origin of magnetism in the human body. The human tissue not only consists of atoms and molecules, but of **elementary magnets** as well. What kind of particles do we mean by that?



Hydrogen protons abound in the body

Atoms consist of a nucleus and electron shells. Hydrogen possesses the least complex nucleus: a single *proton*.

MR tomography uses the magnetic characteristic of the hydrogen protons to generate images.

1. Hydrogen is an elementary part of water and fat, which makes it the most prevalent element in the human body.
2. Of all the elements, the nuclei of hydrogen produce the strongest magnetic resonance signal.

The human body contains *elementary magnets* and is thus *susceptible* to magnetic fields.

Discussed in more detail:

As early as in the 19th century, physicists theorized that 'elementary magnets' might be responsible for magnetism. Today we know: in ferromagnetism, it is electrons, in magnetic resonance, it is atomic nuclei.

Not only protons (hydrogen nuclei) are suitable for use with magnetic resonance. MR imaging and spectroscopy are possible with heavier nuclei as well.

The ability of matter to be magnetized is known as *susceptibility*.

On magnets

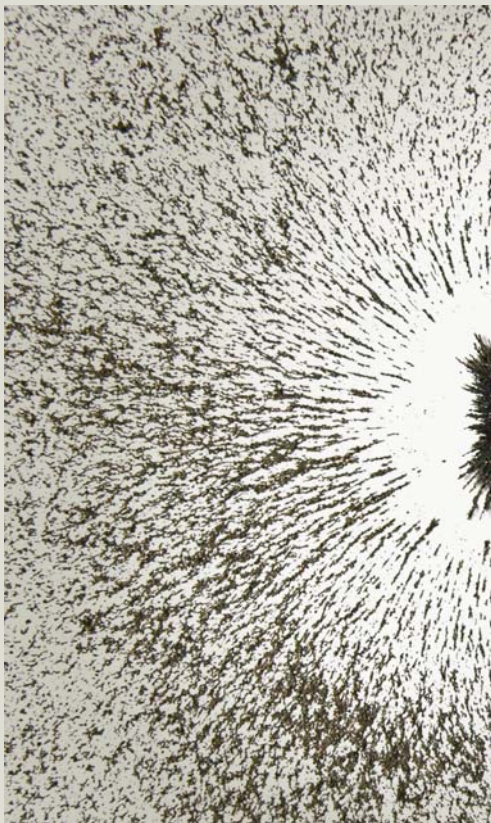
Magnetism is a fundamental property of nature and is seen both in very large and very small structures.

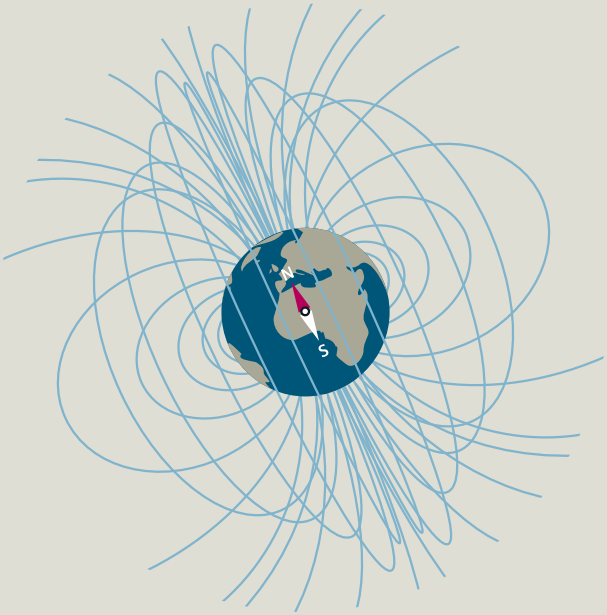
A magnet creates a surrounding **magnetic field**. A magnetic field can also be created by electric currents and electromagnets.

Each magnetic field exerts a force on magnetic and magnetizable particles. Strength and direction of a magnetic field can be visualized by magnetic **field lines**.

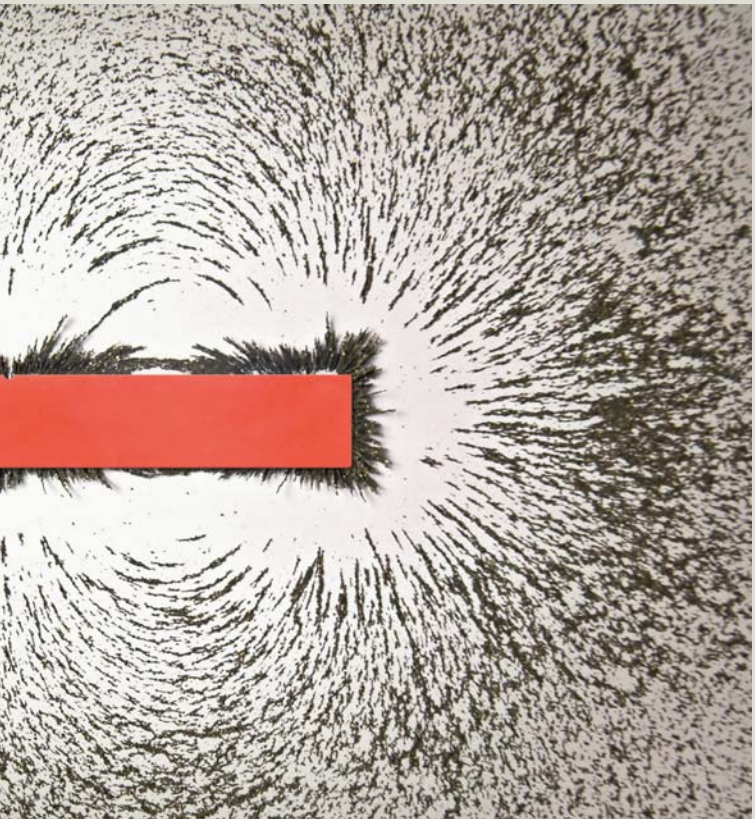
The strength of a magnetic field at each location in space is known as the 'magnetic induction' (symbol: B). In MR technology, the concept of magnetic **field strength** measured in 1 tesla = 10,000 gauss is commonly used. 1 tesla is approximately 20,000 times stronger than the magnetic field of the earth.

A magnetic field of uniform field strength is called a *homogeneous* field. The field lines of a homogeneous field are drawn as equidistant, straight lines running in parallel. A magnetic field that does not change over time is known as a *static* field.





Magnets were knoówn in ancient times in the East and West: the Greek philosopher and mathematician Thales of Miletus (634–546 BC), and the Chinese statesman Guan Zhong (685–643 BC) both made reference to the "loadstone" (magnetite) and its mysterious behavior. With his work *De magnete* (On magnets), British astronomer William Gilbert (1544–1603), brought a scientific understanding to magnetism. Gilbert was the first scientist to realize: "the terrestrial globe is magnetic ... the Earth itself is a giant magnet."



The English physicist Michael Faraday (1791–1867) conceived of magnetic field lines after observing the behavior of iron filings placed on a table between magnets.

The tesla unit is named after the Serbian American physicist and inventor Nikola Tesla (1856–1943).

The proton as a tiny magnet

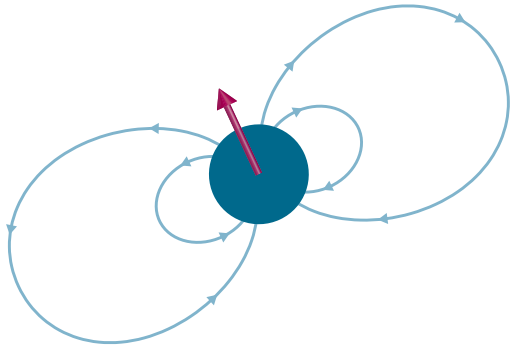
The natural “ingredient” of magnetic resonance

A magnetic moment

We can conceive of the proton as a tiny sphere. As an elementary magnet, the proton possesses a magnetic field similar to the Earth’s magnetic field, with a “north” and a “south” pole.

We attribute to the proton a quantity, known as the **magnetic moment**, which by definition points to “north” (red arrow). The magnetic moment determines the *magnitude* and *direction* of the force this elementary magnet can exert.

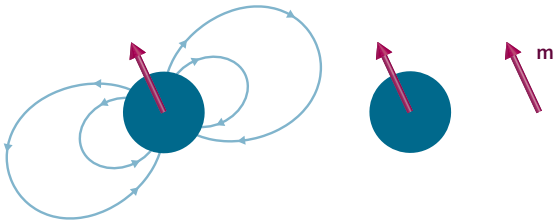
The large number of magnetic protons in the body are responsible for the magnetic reaction of the human tissue to the strong magnetic field of the MR scanner.



The concept of *magnetic moment* was introduced by the French physicist Paul Langevin (1872–1946). He theorized that, at room temperature, molecular magnets display a weak magnetism, also known, in contrast to the strong ferromagnetism, as 'paramagnetism.'

The German physicist Otto Stern (1888–1969) determined the proton magnetic moment in 1933 using the 'molecular beam method' (Nobel prize 1943), further developed by Isidor Rabi for magnetic resonance.

As we are only interested in the proton's magnetic moment (m), we will let its field lines and sphere disappear for good...



Hydrogen protons possess *magnetic moments*, which are responsible for the weak magnetism created in the human body with MRI.

Vectors revisited: arrow arithmetic

As magnetism is a directed physical phenomenon, it might be useful to review the meaning of vectors.

Many physical quantities, for example, temperature or mass, are known to be non-directional. They are sufficiently identified by their magnitude and unit (for example, 70 degrees Fahrenheit, 5 kilograms), so-called *scalars*.

On the other hand, many physical quantities depend on spatial orientation (for example, force or speed). **Vectors** are excellent for defining these quantities, which exhibit *magnitude* and *direction*.

Arrows are suitable for depicting vectors. The direction of the arrow corresponds to the direction of the vector quantity, the length of the arrow corresponds to the magnitude of the vector.

Vector quantities allow for **spatial addition**. The direction has to be taken into account and visualized by linking the arrows.

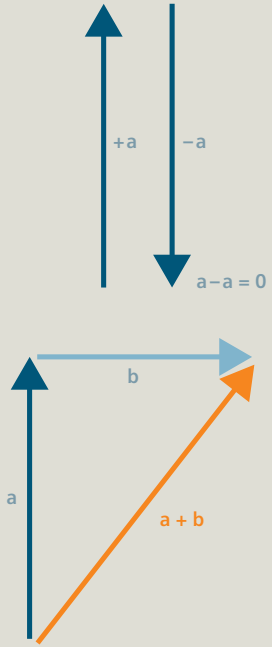
If the arrows point in the same direction, i.e. have the same sense, the magnitude of the vector sum is simply the sum of the magnitudes (in this case **a**+**a**).

Please do not confuse physical quantitites with vectors or vectors with arrows. A vector is a mathematical model for a physical quantity. An arrow is merely a tool for the visual representation of a vector.

Vectors of the same magnitude but opposing direction cancel each other out:
a − **a** = **0**

Just as you can add vectors, you can also decompose them. Each vector, for example, can be divided into separate **components**. These are the projections of the arrow along predefined spatial axes, typically a coordinate system.

In our example, vector c is the sum of vectors **a** + **b**. It consists of a vertical component **a** and a horizontal component **b**.



A different spin on protons

The essence of MRI

About compass needles and spin magnets

When we use a compass, the needle aligns with the Earth's magnetic field and, in principle, points north. The compass needle aligns because:

1. The Earth's field exerts a force on the poles of the needle, which makes it deflect;

2. The needle can rotate freely on a pivot with low friction, although not entirely without friction.

If there were no friction at all, the compass needle would not align but perpetually swing like a pendulum about north.

Exposed to a magnetic field, a hydrogen proton does *not* behave like a compass needle, but more like a spinning top.

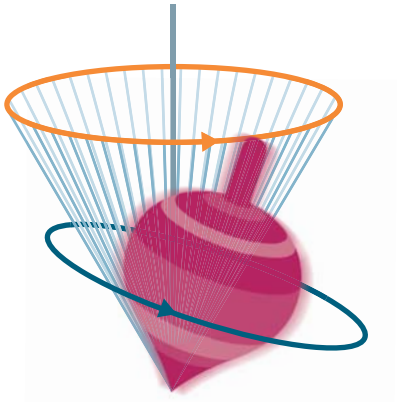
A toy model

As children, we all liked tops. We know that when you tip a rapidly spinning top, it does not fall over. Instead, it begins to wobble because its rotation keeps it from falling over on its side.

This is how the top behaves: its spin axis moves in the shape of a cone about the direction of gravity.

This movement is called **precession**. 'Precession' is the word used by physicists to describe the wobbling movement of a spinning top.

Precession is fundamental to magnetic resonance.

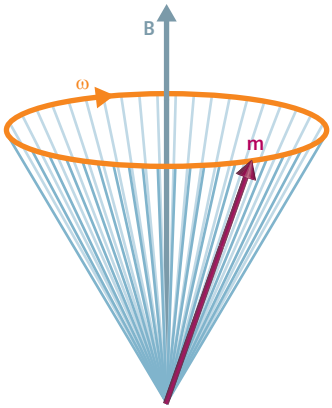


Magnetic tops in the human body

We have learned that a proton does not behave like a compass needle. Let us look at the movement of its magnetic moment *m*) subject to a magnetic field (*B*).

The magnetic moment *cannot* line up with the external field. Just like a child's spinning top, it is forced to *precess*: to follow the shape of a cone around the field. Because the precession of the proton's magnetic moment is frictionless, this movement does not stop as long as the magnetic field is present.

Note that the proton itself is *not* precessing, unlike a child's top. It is rather the proton's magnetic moment, visualized as a rotating vector.

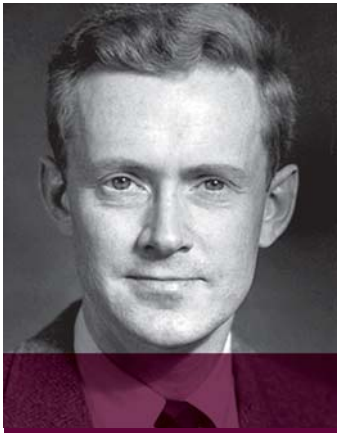


When exposed to a magnetic field, the magnetic moment of a proton *precesses* like a spinning top.

Perpetual spin precession...

As we are surrounded by magnetic fields, elementary magnetic moments always precess, for example, in the Earth's magnetic field. Even though the magnetic field of the Earth is approximately 30,000 times weaker than that of a 1.5 tesla MR magnet, it can, in fact, be used for magnetic resonance.

For clinical imaging, magnetic fields tens of thousands of times the strength of the Earth's magnetic field achieve stronger MR signals and thus, better contrast resolution in a shorter measurement time.



The U.S. physicist Edward Purcell pioneered magnetic resonance in solid matter. "I remember, in the winter of our first experiments, looking on snow with new eyes. There the snow lay around my doorstep—great heaps of protons quietly precessing in the earth's magnetic field." (Nobel prize lecture, 1952)

"(The magnetic dipole) responds like any rapidly spinning gyroscope: instead of lining up with the field, the spin axis precesses around the field direction."

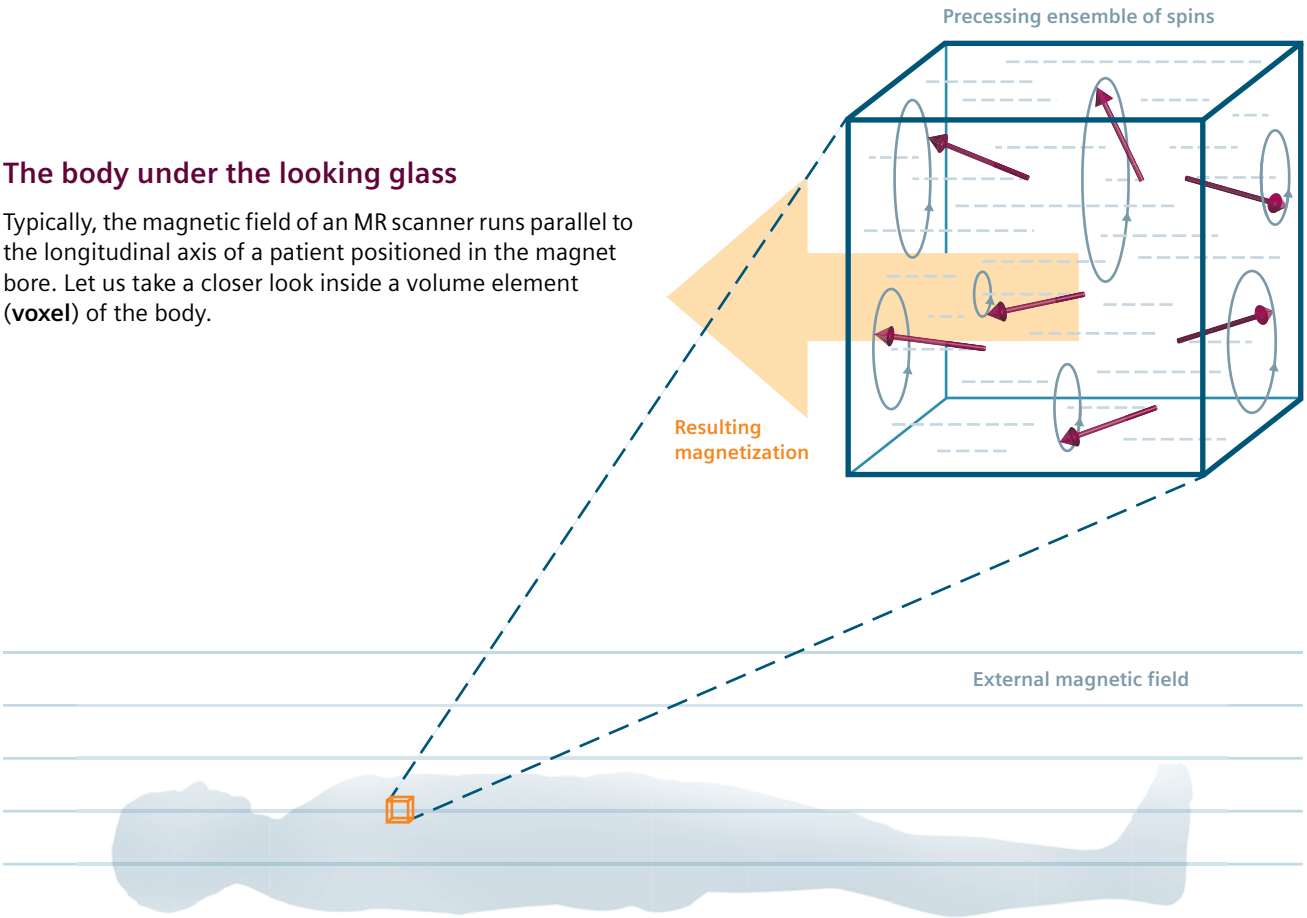
Edward Mills Purcell (1912–1997)

The build-up of magnetization

A wonder of nature exploited by MRI

The body under the looking glass

Typically, the magnetic field of an MR scanner runs parallel to the longitudinal axis of a patient positioned in the magnet bore. Let us take a closer look inside a volume element (**voxel**) of the body.



The total of all proton spins within a voxel constitutes a large statistical number, known as an **ensemble**. As soon as the patient is positioned in the MR scanner, all magnetic moments of the ensemble will precess around the external field lines. They will precess with a *nearly uniform* distribution of directions, but seemingly, a small portion of the ensemble will *tend to align with the external field*.

So, what difference does that make?

It is sometimes believed that nuclear magnetic moments can only align parallel or anti-parallel to a magnetic field. In general, however, the magnetic moments have an *arbitrary* orientation. A static, homogeneous magnetic field *cannot* realign a proton's magnetic moment, only make it *precess*. It is the magnetic and thermal interaction between protons and their *molecular environment* (the 'lattice') that builds up a weak alignment of precessing moments with the external field.

The build-up of longitudinal magnetization takes some time. In a 3 tesla magnet, for example, it will take up to 15 seconds for body fluids, in soft tissue around 4 seconds, for the magnetization to build up. This process is known as "relaxation" and will be explained in Chapter 2, *Echoes, Decay, and Relaxation*.

Partial alignment of spin magnets

A dynamic interaction takes place within the body, partially realigning the precessing magnetic moments. Three "actors" enter the voxel:

1. The ensemble of magnetic moments;
2. The external magnetic field;
3. The molecular environment of the protons.

The interaction of protons with their molecular environment builds up a *macroscopic **magnetization** parallel to the external field* (big arrow in the figure of page 20). We can think of this weak magnetization as the vector sum of the whole ensemble.

Perpendicular to the external field, the magnetic moments *cancel out* because the distribution of precessing moments remains perfectly uniform. Only parallel to the external field does a small excess of magnetic moments remain. This is why the magnetization is purely *longitudinal*.

The larger the number of protons within a voxel (*proton density*), the stronger the longitudinal magnetization. This local magnitude is specific to the type of tissue.

Exposed to a magnetic field, the human body builds up a weak *longitudinal magnetization*, whose magnitude is proportional to proton density.

REDUCED TO THE ESSENTIALS

When the human body is positioned within the magnet bore of an MR scanner, it will acquire a spatial distribution of magnetization parallel to the external field. This longitudinal magnetization is much weaker than ferromagnetim but can be used for magnetic resonance imaging.

The magnetization is caused by the combined effect of hydrogen protons interacting with their molecular environment (relaxation). Hydrogen is the most prevalent element in the human body.

The magnitude of the magnetization is proportional to the number of protons per unit volume (proton density within a voxel). This local magnitude is specific to the type of tissue.

When protons are exposed to a magnetic field, their magnetic moments will precess like spinning tops. Magnetic moments are the vector quantities that determine the magnetic force of the particles.

The beauty of spin

A deeper look into the MR microcosm

The source of magnetic resonance

As we have seen so far, the physical property of a proton relevant for MRI is the vector quantity known as the *magnetic moment*. The reason why the behavior of a proton is different from a compass needle or a bar magnet is explained by the property of **spin**. The spin makes nuclear magnets precess like spinning tops.

How can we picture the property of spin? In fact, the spin of a proton cannot be attributed to intrinsic rotation. As an analogy, we can visualize the spin as something like a spinning wheel. A magnetic spinning wheel.

Let the spinning wheel tilt and wobble, and we have precession.

The speed, or frequency, of precession is of great importance in magnetic resonance: We excite many proton spins to tilt and precess "in phase," and thus generate an MR signal.

In 1921, the American physicist Arthur H. Compton (Nobel prize 1926) supposed “that the electron is spinning like a gyroscope (and) possesses a magnetic moment.” It was not until 1926 that Dutch physicists Uhlenbeck and Goudsmit reestablished the idea of “self-rotation” of particles in order to explain the hyperfine splitting of spectral lines. In terms of physics, the notion of self-rotation of subatomic particles cannot be upheld, but the word “spin” has been coined and is here to stay.

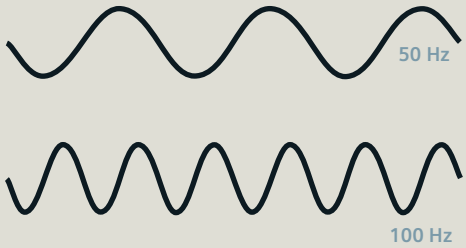
On frequencies and phases

What do we mean by **frequency**? It is the number of revolutions or oscillations of a periodic movement per unit time.

You know this from your car and the tachometer. For example, the tachometer shows 3,000 revolutions per minute. This is nothing more than the frequency.

3,000 rpm are the same as 50 revolutions per second. The unit for revolutions per second is **hertz** (Hz). In our case, this means a frequency of 50 Hz.

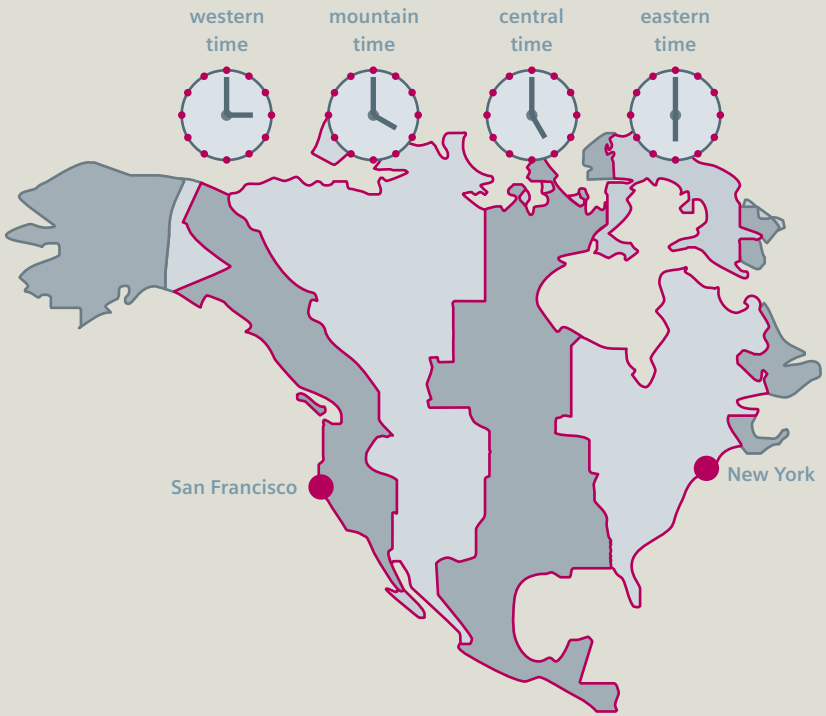
When we apply oscillations to a time axis we obtain the peaks and crests of a sinusoidal wave. Oscillation with double the frequency (100 Hz) is shown as a compressed sinusoidal curve.



You can compare a **phase** to the movements of the hands on a watch or clock. The hands show you the offset in time between one state of rotation and another.

If your watch is an hour fast, you could think of this as a "phase shift" of 1 hour relative to local time. You could correct this by setting your watch accordingly, that is, by setting the hour hand back by 30 degrees. This is not possible when we talk about the phase shift between San Francisco and New York. The three-hour time difference between the two cities is of a lasting nature. If you travel in a plane across large distances, you can experience this kind of time shift as jet lag. Most oscillations, for example, radio waves, contain this type of "jet lag."

Later, we will describe how both shifts in frequency and phase are used to compute images from the MR signals acquired.



Precession revisited

Proportional to field strength

The Larmor relationship

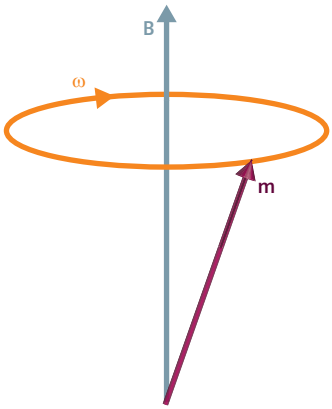
The precession frequency of spins is also known as the **Larmor frequency**. It depends on the type of nucleus and the strength of the magnetic field applied.

The Larmor frequency ω increases proportionally with the magnetic field B. The following expression applies:

$$\omega = \gamma B$$

(The constant factor γ is known as the gyromagnetic ratio of the nuclei.)

In the earth's magnetic field, spins precess relatively slowly at approximately 2,000 Hz (2 kHz).

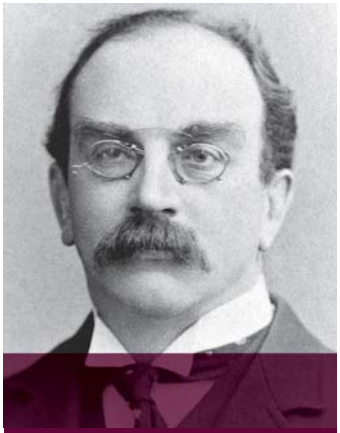


The quintessential movement in MRI

In the magnetic field of an MRI scanner, spins will precess at frequencies comparable to *radio frequencies*. This means that the spins precess at several million revolutions per second. The Larmor frequency in a 1.5-tesla magnetic field is 64 MHz, in a 3-tesla field it is twice that, i.e. 128 MHz.

Larmor precession is at the heart of magnetic resonance. It is the physically most adequate description of spin dynamics in magnetic fields.

Larmor precession of spins is *proportional* to magnetic field strength.



"The influence of a magnetic field on the motions ... is precisely the same as that of a rotation with angular velocity ω ... around the axis of the field."

Joseph Larmor
(1857–1942)

Spin is a quantity that adds to the total angular momentum of a particle. In this book, we don't look at spin as such, which is a complex object residing in a cone-like probability cloud, describable by quantum physics. In MRI, discussion of spin is reduced to observable components of spin along a quantizing axis ("spin up", "spin down"). The relevant aspect of spin for MRI is the averaged value (expectation value) of the spin magnetic moment, visualized as the *spin polarization vector*, also known as the "Bloch vector." This vector is the precessing quantity shown. And this is what, in MRI, is meant by 'spin' for short.

Nuclear spin beyond protons

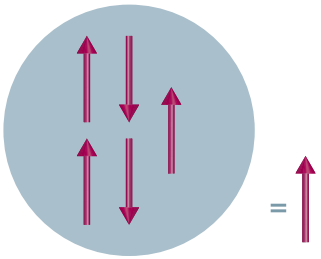
Sources for magnetic resonance imaging

Nuclei with net spin

MRI predominantly uses hydrogen nuclei (protons). Atomic nuclei in general consist of protons and neutrons. Both have the property of spin.

Nuclei with an *uneven* number of protons or neutrons have a net spin known as the **nuclear spin**. Common examples are: carbon ^{13}C, fluorine ^{19}F, sodium ^{23}Na, or phosphorus ^{31}P. Two thirds of the isotopes found in nature have a net nuclear spin, making them suitable for use with magnetic resonance.

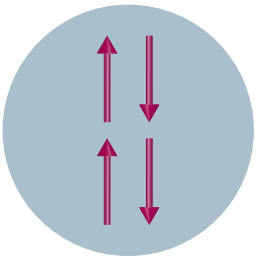
As explained before, the Larmor frequency is not only proportional to the external field strength but depends on the type of nucleus as well (identified by the gyromagnetic ratio). Exposed to a 1.5-tesla field, hydrogen protons precess with a different Larmor frequency than, for example, phos-phorus nuclei.



Nuclei without spin

Nuclei with an *even* number of protons and neutrons do *not* have a net nuclear spin. They are magnetically neutral. Examples of these are oxygen ^{16}O (with 8 protons and 8 neutrons each) or carbon ^{12}C (with 6 protons and 6 neutrons each). These isotopes are *not suitable* for use in mag-netic resonance.

Atomic nuclei with an *uneven* number of protons or neutrons have a *net spin* and are suitable for magnetic resonance imaging.



= 0

In the atomic nucleus, two identical particles cannot be in the same state. They have to align their spin *anti-parallel* to each other, and the net spin of this "couple" of particles cancels out. This rule of nature is known as the *Pauli Verbot* (or 'exclusion principle'). It is the residual particle that creates the nuclear spin. The presence of nuclear spin *does not* mean that the nucleus is spinning but has a net magnetic moment.

The small difference that makes MRI possible

Focusing on the “suspects” that are responsible for the magnetization

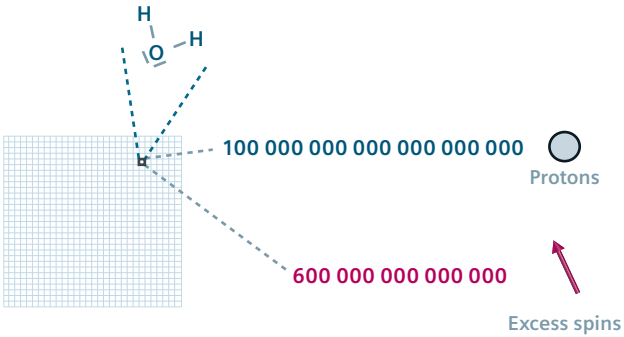
Introducing the concept of excess spins

Magnetization is built up in a spin ensemble because the ratio of spins precessing in and against the direction of the external magnetic field is not 50:50 but slightly biased. There are slightly more spins pointing in the direction of the exter-nal field than against the field. We ignore the majority of spins that cancel each other out and focus solely on these **excess spins** : the relatively small number of spins consid-ered to add up to longitudinal magnetization.

The number of excess spins is a function of different factors. Their number grows:

- in proportion to the **proton density**
- with the strength of the external magnetic field
- as the temperature decreases.

At body temperature and a field strength of 1 tesla, there are approximately 6 excess spins, or 0.0006 % of 1 million pro-tons in a voxel of water.



The excess, or population difference, of precessing spins in a magnetized volume is expressed by the well-known Boltzmann distribution of statistical physics.

Calculating the excess

Since it takes many zeros after the period to express it in percentages, there is another way of expressing this small numerical relationship. The unit of measurement for very small quantities is ppm or **parts per million**. Coming back to our example, at 1 tesla, the number of excess spins is approximately 6 ppm.

As you can see, the number of excess spins is relatively small. That we are still able to obtain a measurable effect is due to the large number of protons in the human body.

For example: a tiny voxel of 1.5 microliters of water is populated by approximately 10^{20} hydrogen protons (a 1 followed by 20 zeros). At 1 tesla with 6 ppm excess spins, this translates into 600 trillion small spin magnets that add up to a macroscopic magnetization.

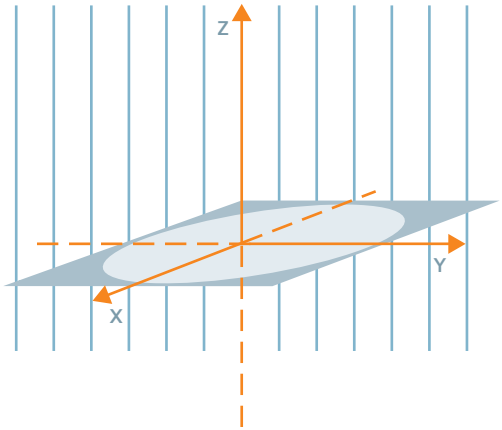
Excess spins are the small number of spins within a voxel that add up to net magnetization.

The xy-plane

Let us agree on the following conventions:

In the usual xyz coordinate system, we place the **z-axis** by definition in the direction of the external magnetic field.

We will call the plane running transverse to the field lines the **xy-plane**.

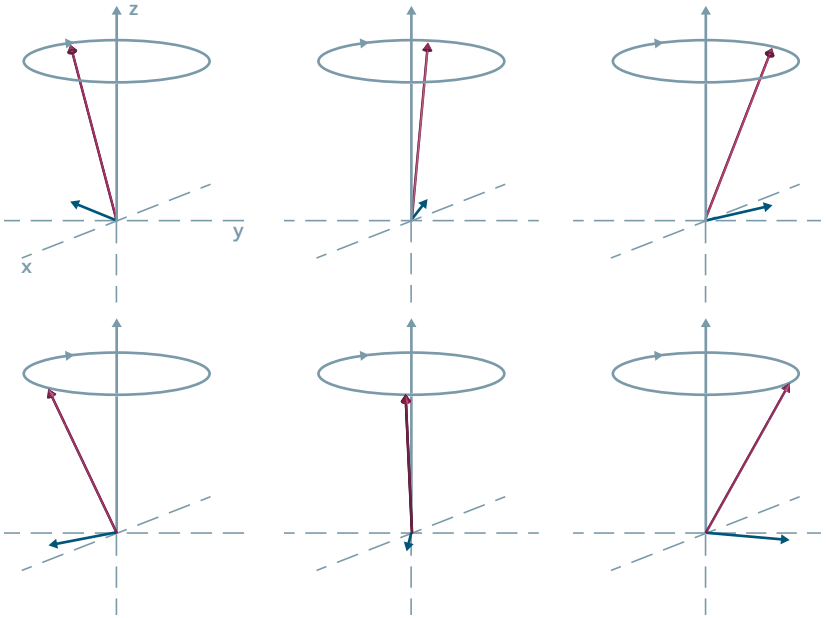


The spin packet

We are going to concentrate on the excess spins of an ensemble. Let us reduce them for clarity's sake to a spinning "six pack", with identical precession cones but different phases.

While all spins of the spin packet have the same precession frequency, they point in different directions. The components transverse to the magnetic field (blue arrows), that is, within the xy-plane, cancel out to zero.

Consequently, magnetization will be built up along the z-axis only, as a statistical sum of the z components of the spin vectors. This is the *longitudinal* magnetization.



REDUCED1 TO THE ESSENTIALS

The precession of magnetic moments of nuclei is due to its spin property. We can picture spin as a kind of magnetic spinning wheel.

Along with hydrogen, two thirds of the isotopes found in nature have a net spin, making them suitable for use in magnetic resonance.

The velocity of precession, known as the Larmor frequency, is a function of the magnetic field strength applied. In MRI, we concern ourselves with field strengths that correspond to the high-frequency range of radio waves (in Megahertz).

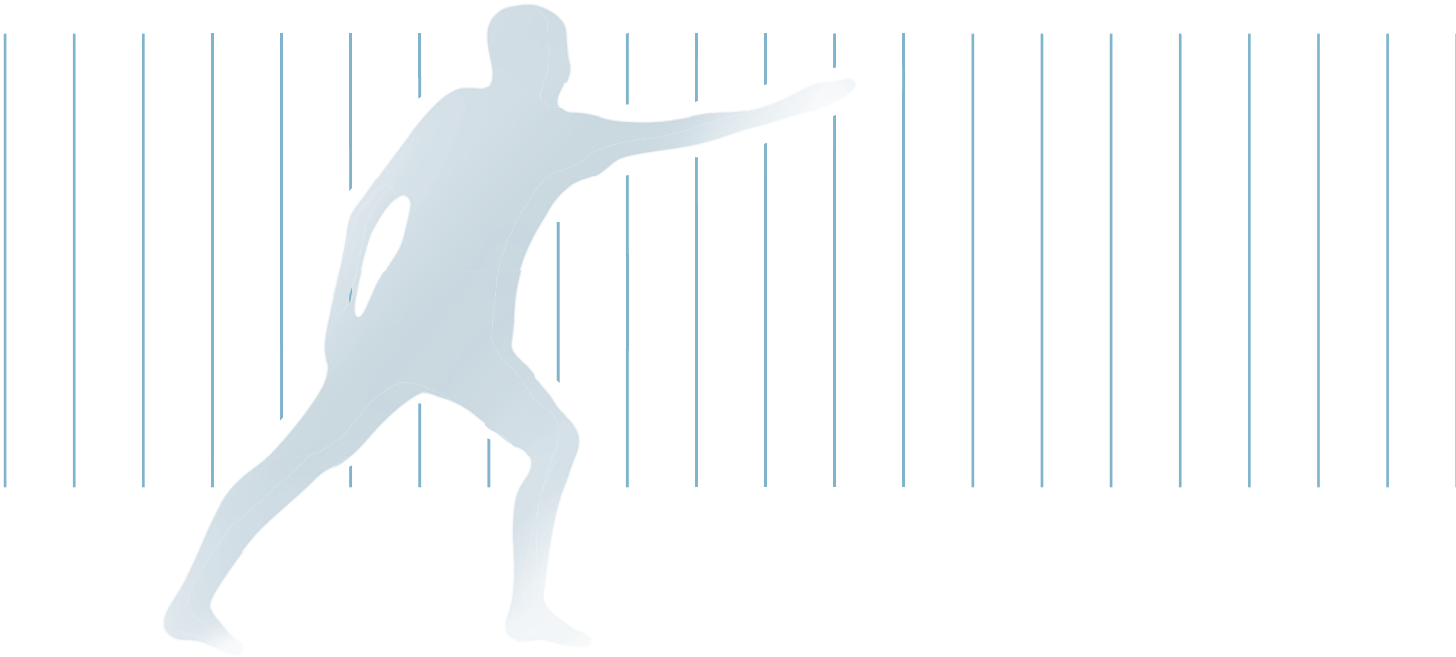
A spin packet is a model for the excess of spins precessing at the same frequency in a voxel, which are considered to add up to macroscopic magnetization.

Creating the resonance

Making spins generate the MR signal

The excess spins within a voxel create a static longitudinal magnetization. In order to generate the MR signal, we tilt the magnetization so that a component will precess in the xy-plane.

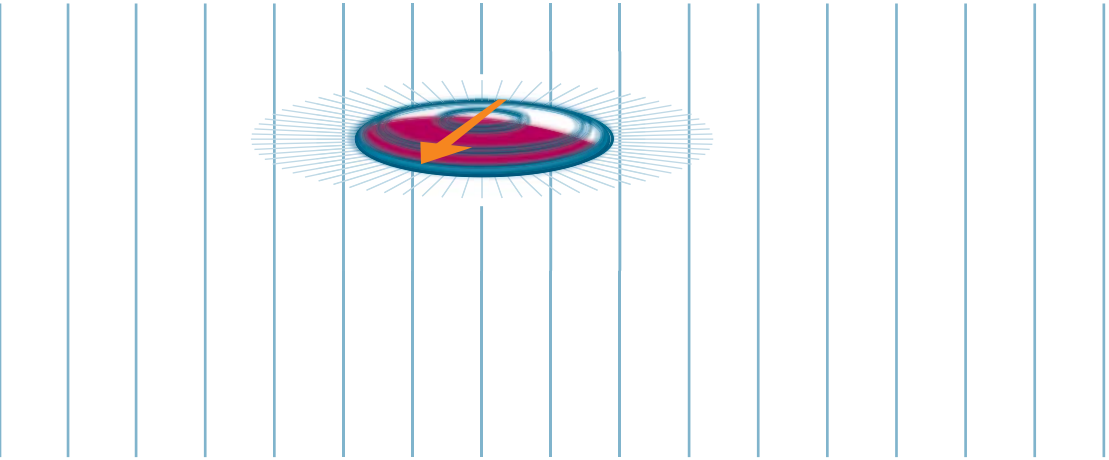
The magnetization can be tilted by applying a short electromagnetic pulse, the **RF pulse**.

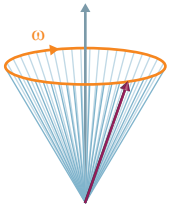


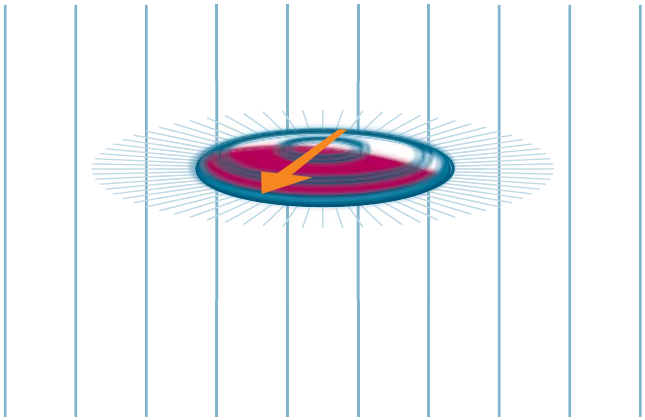
The RF pulses transmitted during a patient examination use a *circularly polarized wave* that contains a rotating magnetic field.

Magnetic frisbees

What is an RF pulse? Imagine a magnetic frisbee that is suddenly flying through the field. The frisbee acts like a *rotating magnet*. How does a rotating magnetic field tilt the magnetization?







Satisfying the resonance condition

For maximum impact, the RF pulse acts perpendicular to the external field and is in **resonance** with the precessing spin packet of the voxel of interest.

Physically, the *resonance condition* means:

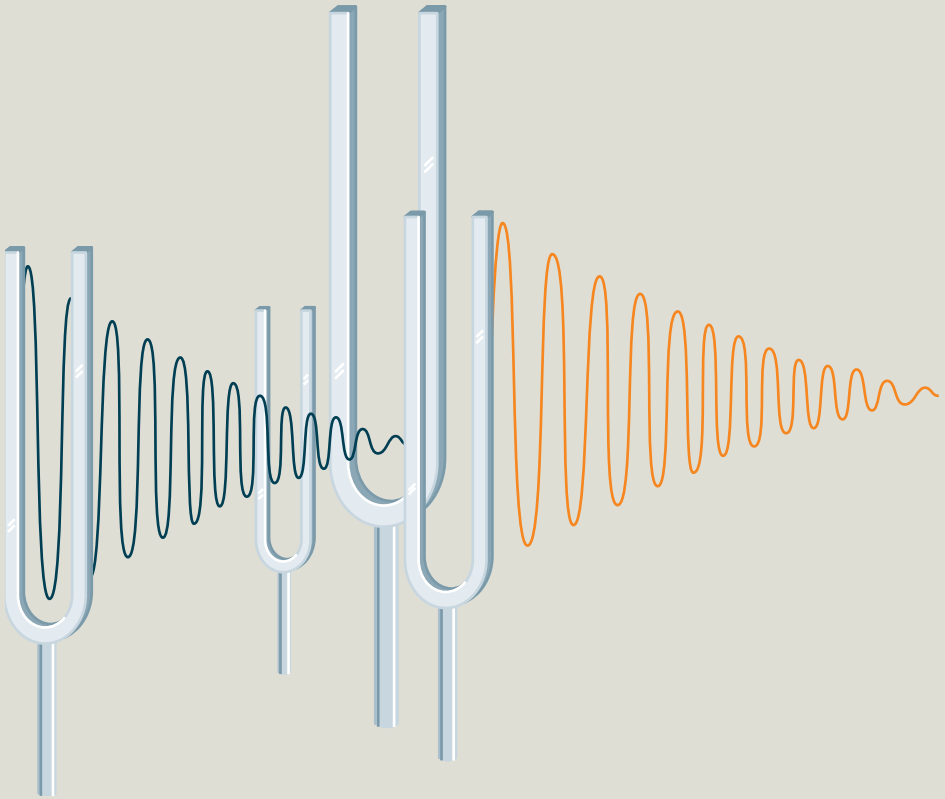
The frequency of the RF pulse applied has to match the Larmor frequency of the spins.

Returning to our analogy: the rotating magnet has to rotate at the same speed as the spins precess.

On resonances

Resonance stimulation in MR can be compared to the oscillations created by a tuning fork. When the tuning fork is struck, it begins to oscillate and generates a specific sound. The pitch corresponds to the oscillation frequency of the acoustic wave.

When you introduce a second tuning fork tuned to the same frequency, it will oscillate in response to the acoustic waves emitted by the first tuning fork. The two tuning forks are now in resonance.



Tilting the magnetization

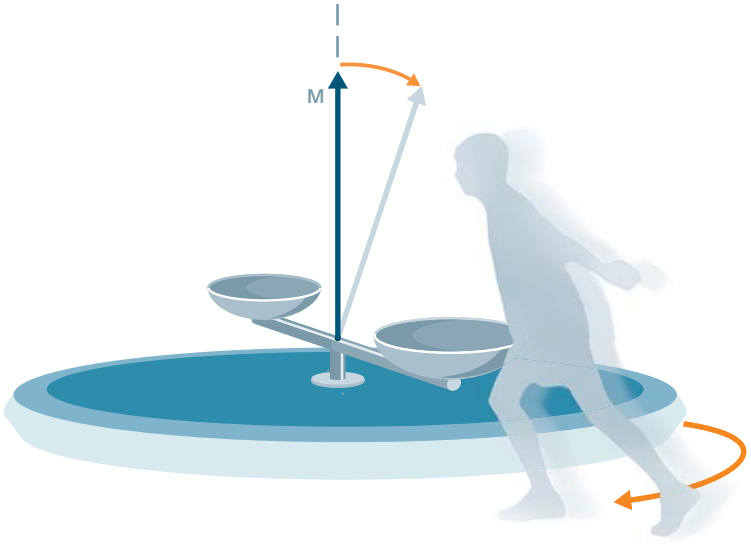
A rotating frame

The story of the merry-go-round

What exactly is happening during magnetic resonance? Let us look at yet another analogy:

Imagine yourself as the rotating magnet (that is the RF pulse) running in *resonance* with the precessing spins.

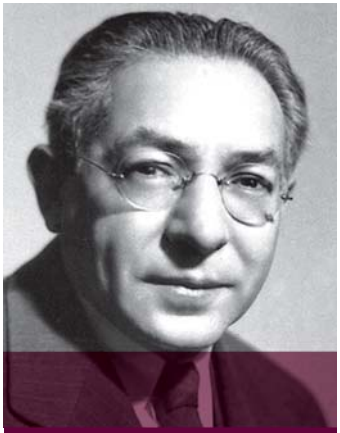
For this purpose, you are running around the spin carousel throwing stones into a rotating "spin scale." Your time is limited. In you run either too fast or too slow around the merry-go-round, you are no longer in step. Now, you can only catch up with the scale after one complete revolution and you can only throw one stone. But if you are in step with the spin scale, you can throw as many stones as you want into the scale.



As a result of your efforts, the spin scale loses its equilibrium and its axis simply flips over.

This is exactly what the RF pulse does: it tilts the magnetization (M).

In magnetic resonance, the 'merry-go-round' is known as the **rotating frame** of refer-ence, introduced by Isidor Rabi and Felix Bloch.



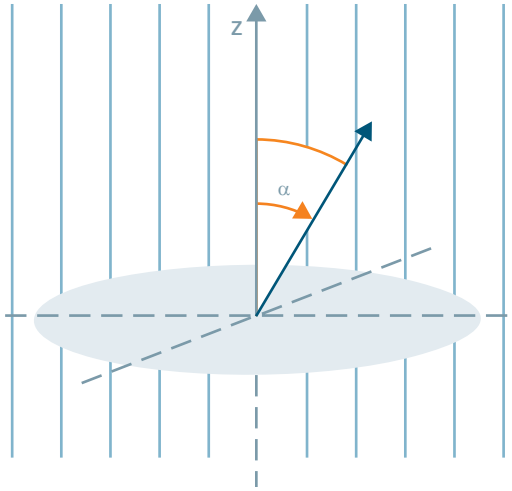
"To the system ... we apply an additional magnetic field which is much smaller than H_0 and perpendicular to it in direction"

Isidor I. Rabi
(1898–1988)

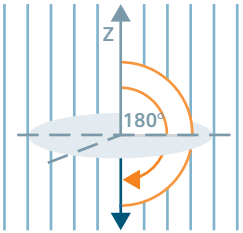
The American physicist Isidor Rabi (Nobel prize 1944) discovered magnetic resonance in 1938, using the technique of molecular beams. The tilting of spins around the field axis of the RF pulse, perpendicular to Larmor precession, is known as 'Rabi precession.'
"After the first experiment, every-thing was easy. Since you're doing experiments of that sort, there ought to be something aesthetic about them."

Pulses and flip angles

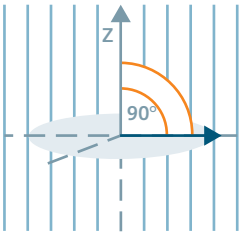
The stronger the energy of the RF pulse, the farther the magnetization will flip or tilt. The final tilt angle is known as the **flip angle** (α).



A **180-degree pulse** flips the magnetization into the opposite direction of the z-axis.



A **90-degree pulse** flips the magnetization exactly into the xy-plane.



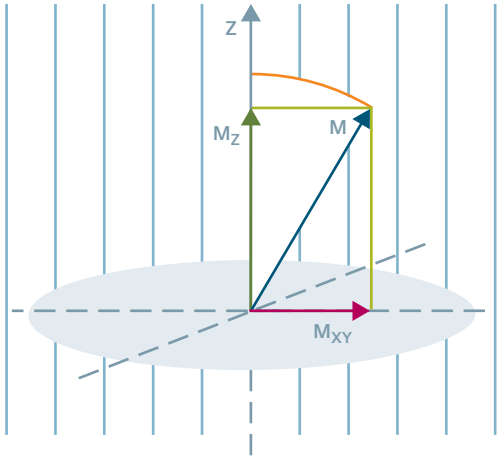
Decomposing the magnetization

Just like a vector, the tilted magnetization can be decomposed into two components located perpendicular to each other:

Longitudinal magnetization M_z is the portion of the vector in the z-direction, that is, along the external magnetic field.

Transverse magnetization M_{xy} is the component of the vector that rotates about the external field in the xy-plane. How fast does it rotate? The rotating transverse magnetization is the sum of the spins that rotate in phase in the xy-plane, matching the Larmor frequency. Therefore, transverse magnetization also rotates with the Larmor frequency.

After magnetic resonance, the resulting transverse magnetization will rotate with Larmor frequency.



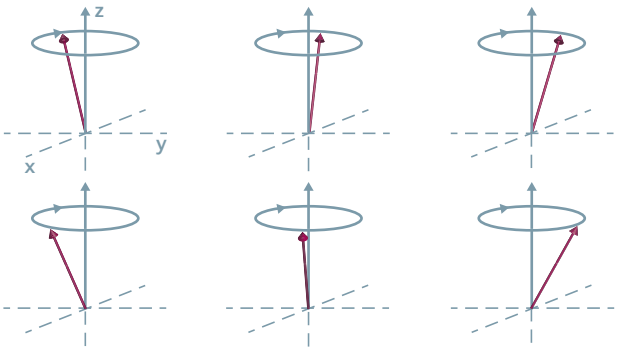
A closer look at the resonance situation

The viewpoint of the spin packet

By applying a **90-degree pulse**, magnetization flips in the transverse direction, that is, in the xy-plane. As long as the RF pulse is present, two magnetic fields are active: the static field, and for a short time, the rotating RF field.

Using a special trick, we can make the static field disappear: all we need to do is climb onto the spin carousel (the rotating frame) together with the spins. Once on it, the spins effectively "feel" only the rotating RF field (the frisbee magnet). Since this field rotates in resonance with the spin packet, its axis appears *static* to the spins.

How do the spins react to this magnetic excitation?

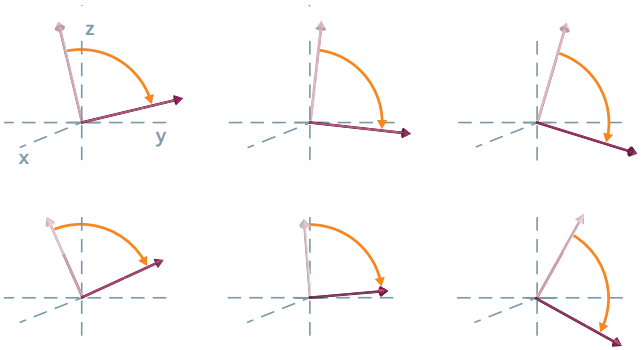


Before the 90-degree pulse

The spin packet starts to precess "downward" to the xy-plane. After a 90-degree pulse, they are concentrated in the horizontal direction (to the right in our example).

As you can see from our picture, the xy components of the spin packet do not point randomly in all directions any more, instead they all share more or less one direction. Now they precess *coherently* and thus generate a transverse magnetization.

As you can imagine, when a **180-degree pulse** is applied, the spin packet will rotate until it is flipped downward by 180 degrees. Accordingly, longitudinal magnetization will be oriented antiparallel to the external field.



At the end of the 90-degree pulse

Here comes the MR signal

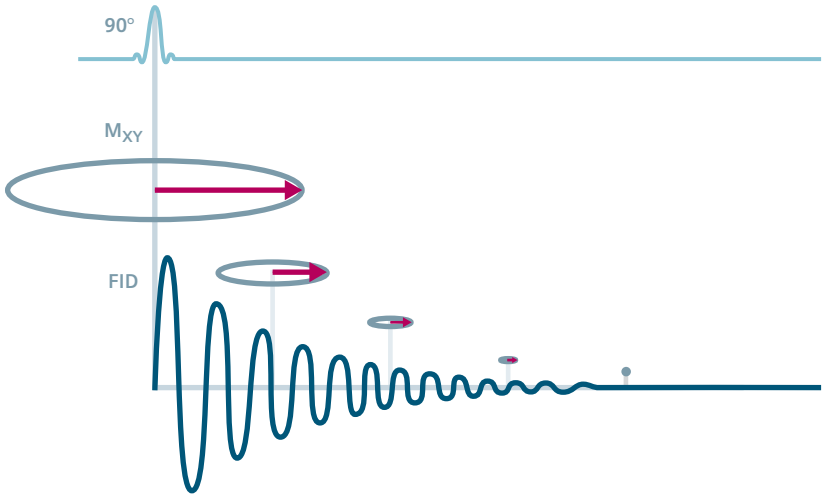
Oscillating with Larmor frequency

Rotating transverse magnetization

The transverse magnetization M_{xy}, subject to the static magnetic field, behaves like a rotating magnet. You can move a coil into the rotating magnet and induce an electric voltage in it.

The course of the voltage over time is the **MR signal**. The stronger the transverse magnetization within a voxel, the stronger the MR signal.

By acquiring different signals from different voxels, image contrast is created.



Free induction decay

Note that the MR signal decays relatively quickly. Since transverse magnetization

- precesses *freely*
- *induces* a signal, and
- *decays* immediately

after the end of the RF pulse, this MR signal is called **free induction decay** or **FID** for short.

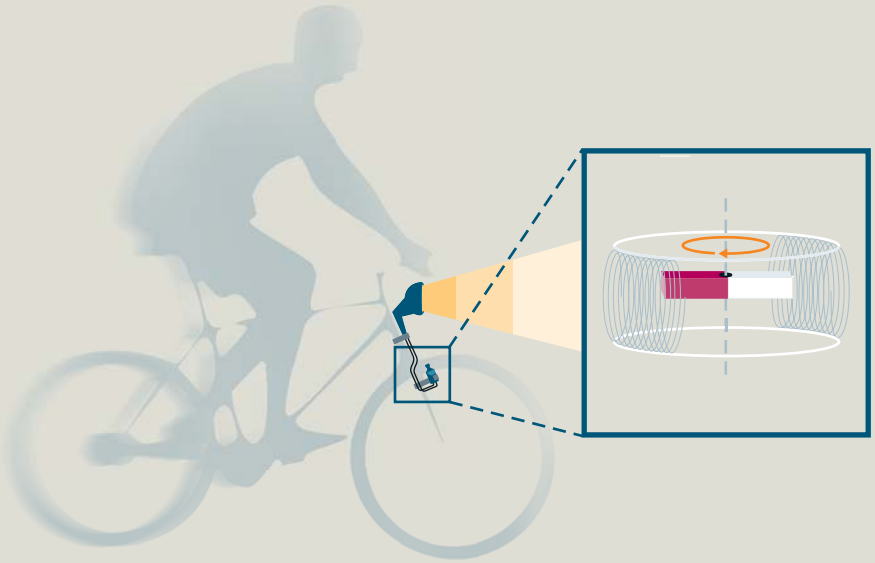
As the transverse magnetization does not rotate forever but starts to decay, the induced signal will decay, too.

The reasons for the decay of the MR signal are explained in Chapter 2, *Echoes, Decay, and Relaxation*.

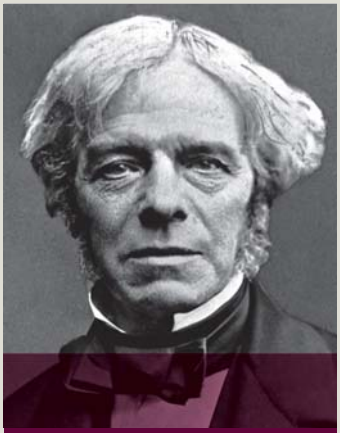
On induction

From electrical engineering we know that a magnetic field changing in strength or direction generates an electric voltage in a coil. This is what we call **electromagnetic induction**.

We use this induction every single day. For example, a bicycle dynamo contains a rotating magnet driven by the wheel of the bicycle which continuously changes the direction of its magnetic field.



These changes in magnetic field induce a flow of current in the coil, which can be used to light up the bicycle lamp. The faster the dynamo magnet rotates, the higher the induced electric voltage and the brighter the light generated by the bicycle



"If a single wire be moved like the spoke of a wheel near a magnetic pole, a current of electricity is determined through it ..."

Michael Faraday
(1791–1867)

Electromagnetic induction is a near-field effect and can be described by classical electrodynamics. Sometimes the MR signal is believed to be a radio-frequency wave, which is received by the RF coil like a radio signal. But in MRI, the radio wave of the spins is negligible. The Faraday field at the coil is much larger and is responsible for the signal detected.

Michael Faraday not only explained magnetic field lines but also discovered the phenomenon of electromagnetic induction in 1831, also known as "Faraday induction" in honor of his name.

SUMMARY

In a strong magnetic field, a locally varying distribution of magnetization is created in the human tissue.

The external field forces the proton spins to precess at the Larmor frequency, which is proportional to the strength of the magnetic field.

The molecular environment forces the spins to realign, breaking the uniform distribution along the external field. A weak magnetization is built up parallel to the external field.

When excited by a 90-degree RF pulse, longitudinal magnetization is tilted into the xy-plane and rotates at the Larmor frequency.

The rotating transverse magnetization generates the MR signal in a receiver coil and decays quickly (free induction decay, FID).

The technical side

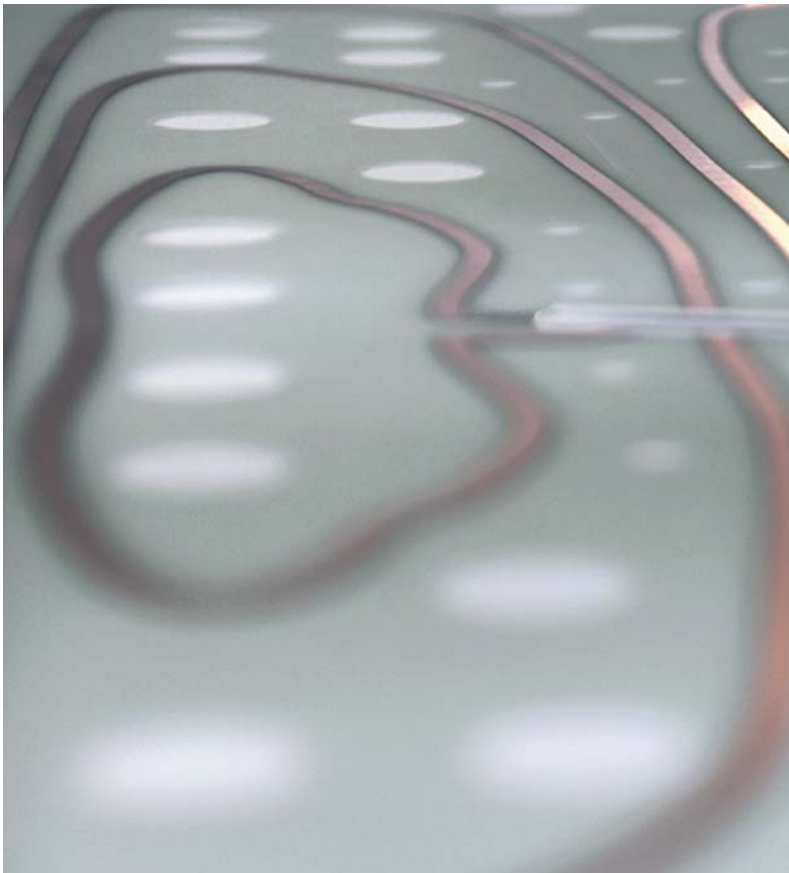
The components that make MRI possible

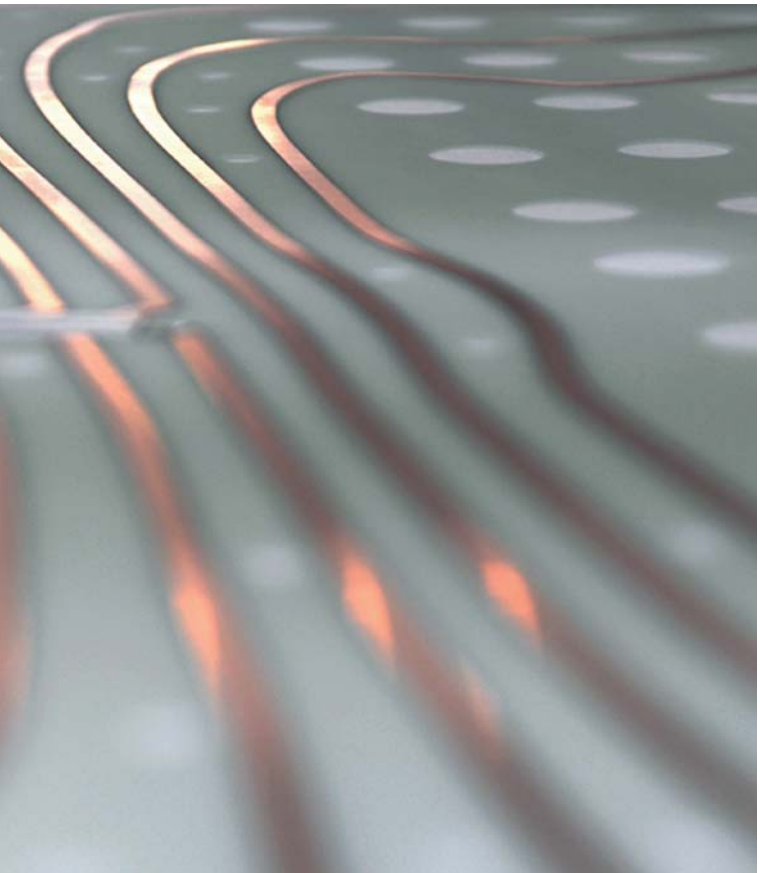
The main system components

A typical MR system consists of three components or subsystems:

- a magnet with a main magnetic field,
- a gradient system, and
- the radio-frequency (RF) system.

The gradient system is needed predominantly for the localization of slices and voxels and will be explained in Chapter 3, *From the Signal to the Image*.





The computer system

To generate and evaluate high-quality MR images, the three subsystems have to be controlled and the measured results displayed. For this purpose, the high-performance computer system includes the following:

- the image processor for computing the MR images from raw data acquired
- the host computer with console for the technician, and
- the control as well as evaluation software for performing measurements and creating image results of diagnostic value

The main magnet

A strong magnet

The homogeneous magnetic field required for MR imaging is generated by a strong magnet. This magnet constitutes the most important as well as the most expensive component of the MR system.

In general, superconducting electromagnets are used. A strong magnetic field is generated by the electric current flowing in large coils.

The homogeneous magnetic field is located in the center of the magnet bore, which runs parallel to the longitudinal body axis of the patient.

Superconducting magnets

This is how a superconducting magnet works: At normal temperature levels, an electric conductor is resistive. Without a constant power supply, an electric current injected into a circuit would begin to decay because of its loss in energy.

Superconductors are materials that have no electric resistivity at very low temperatures close to absolute zero (0 Kelvin = – 273 °C). A constant, high current (above 400 amperes) will flow for years without an electrical poten-tial or voltage. For this reason, the superconductor has to be kept at very cold temperatures. Liquid helium is used as a coolant.

The RF system

The radio-frequency (RF) system

The proton spins of the body tissue are excited by briefly applied alternating magnetic fields in the radio-frequency range. These RF pulses are transmitted, the MR signal generated by the rotating transverse magnetization has to be received.

The RF system of an MR installation consists of the following:

- RF coils
- RF transmit amplifier for sending RF pulses
- RF receive amplifier for amplifying the received MR signal

RF coils

The body transmit coil is integrated in the magnet bore. Local RF receiver coils come in all shapes and sizes. The shape of the coil is determined by its area of application. Depending on the body region to be examined, they are positioned locally on the patient's body.

The better a local coil is adapted to the geometry of the human body, the less unwanted signal from surrounding tissue can influence the imaging process.

Array coils are used for examining larger measurement areas. They include several independent smaller coil elements that can be combined according to the area under examination.

The magnetic field: safety aspects

Biological effects

The strong magnetic field of the MR scanner affects the tissue as well as all other magnetizable material in the vicinity of the magnet.

Since the introduction of MR tomography, a number of studies have been performed to determine the biological effects of the static magnetic field. Known effects include, for example, dizziness, stomach upsets as well as a metallic taste. Most of these effects only occur at field strengths above 3 tesla. These are *short-term effects*, that is, they occur exclusively in the magnetic field or shortly after leaving it. To date, no biological long-term effects have been observed.

The distribution of surface currents present during an ECG alters in the magnetic field (magnetohydrodynamic effects). Cardiac functions are not affected by it, only the observed ECG signal is.

Magnetic effects on devices and material

Magnetizable materials, for example iron, are attracted by the MR magnet. This constitutes a potential source of hazard to the patient or the operating personnel. Considerable forces may be generated, attracting even large iron masses and causing them to accelerate as they move toward the magnet. The force exercised is proportional to the mass involved.

Metal parts in the patient are also a source for hazard. Metal splinters, clips, screws, or injection needles may be moved in the body by the magnetic forces.

Especially critical are electrical implants such as pacemakers or hearing aids. As determined by national and international recommendations and guidelines, the safety/exclusion zone for pacemakers has been established at a field strength of 0.5 mT outside the magnet.

The functionality of hearing aids may be compromized in strong magnetic fields.

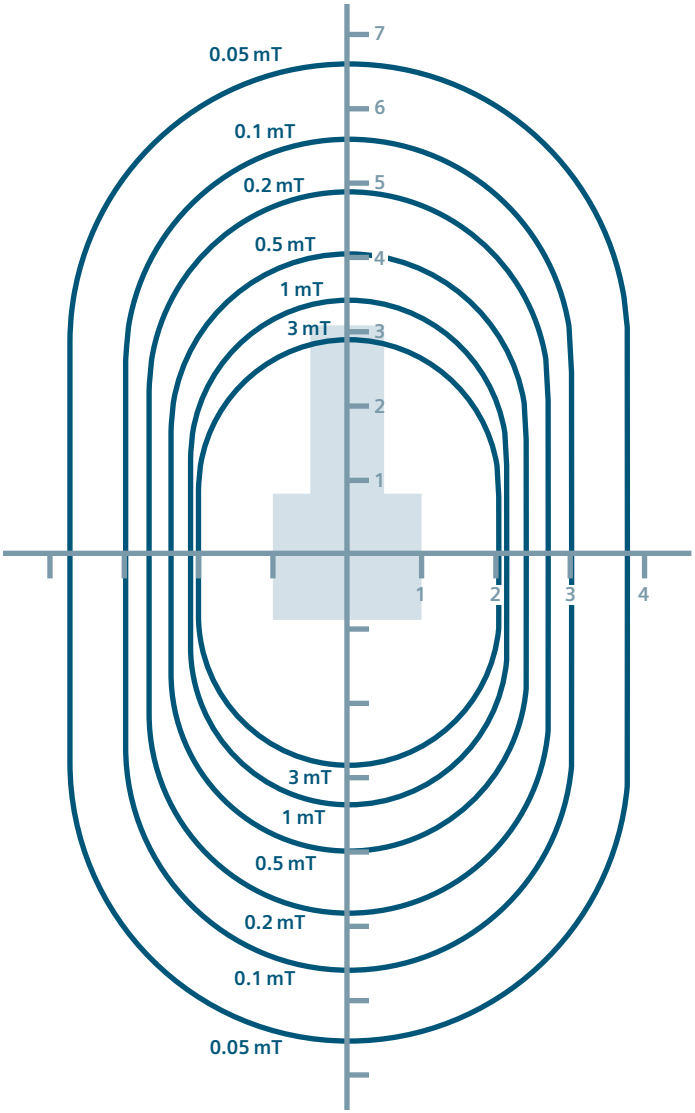
In each case, the patient must be interviewed prior to the actual MR examination. If there are reasonable doubts, other examination methods should be used.

The functionality of mechanical devices and electric components is *not* ensured in the vicinity of the magnet. The functions of clocks, respiratory devices, as well as monitors, infusion pumps, and other devices may be affected by the magnetic fringe field. The same applies to computers and magnetic data carriers. Also, the encoding on credit cards may be deleted in the vicinity of the magnet.

Effect of the fringe field

The typical field strengths of magnets used in today's whole-body MR tomographs are up to 3 tesla and may exceed 7 tesla in special cases. The MR magnets not only generate the desired nominal field in the area under examination, they also generate a **fringe field** outside the magnet.

The strength of the fringe field as well as its spatial distribution depends on the configuration of the magnet, its size, as well as its basic field strength.



Shielding the fringe field

For super-conducting magnets, the fringe field is shielded by using measures that limit the external safety zone.

Today, **active shielding** is used. The fringe field is largely compensated by additional super-conducting coils wound in the opposite direction on the field generating coils.

The RF field: safety aspects

Tissue warming

RF electromagnetic waves generate currents in electrically conducting tissue and stimulate molecules in the tissue. The resulting oscillations lead to tissue warming. Usually the increase in temperature is less than 1 degree Celsius.

The **specific absorption rate** (SAR) is the RF output absorbed per time unit and kilogram.

For safety reasons, the RF power emitted by the system into the body is monitored and the respective SAR values are limited accordingly. The IEC limit values are 4 W/kg (whole body), 4–10 W/kg (partial body), 20 W/kg (local SAR head, trunk), and 40 W/kg (local SAR extremities).

If the receiving RF coil is in resonance with the transmitter, it may act to increase the RF field close to the coil. This increase in field strength is of particular concern when it occurs near the eyes. To eliminate this effect, the system decouples the receiver coil during transmission.

The RF field may induce AC currents in metal implants or cables routed close to the patient (for example, ECG cables), resulting in local warming.

System-specific warnings, labels, or notices have to be observed at all times.

Interference caused by other systems

The RF field emitted by the transmit coils may be superimposed on the voltage in external devices and lead to interference.

Conversely, external interference (for example from radios, cell phones, electronic controls, electro motors) may emit interfering signals into the MR system and degrade the image quality.

To provide the best possible protection in both directions, MR systems are installed in RF-sealed rooms made from conducting materials (Faraday cages).

Echoes, Decay, and Relaxation

After RF excitation, the rotating transverse magnetization decreases rather quickly and the MR signal decays. We can create echoes of the lost MR signal for a while. It takes some time before the longitudinal magnetization fully recovers. These processes are known as relaxation.

A first acquaintance with relaxation

Two seemingly independent processes

Decay and recovery

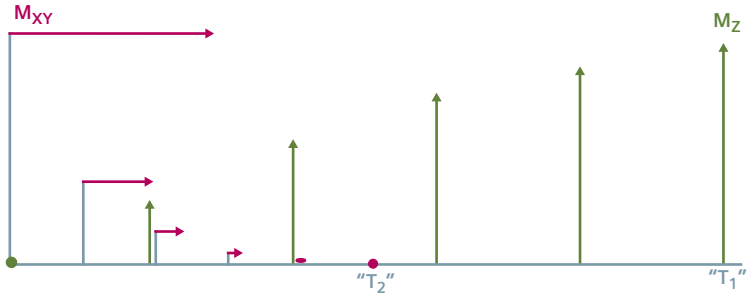
After the longitudinal magnetization has been tilted in the xy-plane by an alternating field (RF pulse), it immediately rotates at Larmor frequency and induces the MR signal (FID).

Two things happen:

1. Transverse magnetization M_{xy} (red) decays.
2. Longitudinal magnetization M_z (green) recovers.

You might think that the following would occur: the magnetization vector simply tilts back to its original direction parallel to the external field.

However, this is not true.



Two additional contrast mechanisms

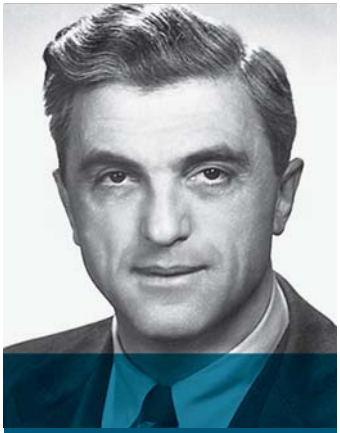
Transverse magnetization M_{xy} decays more rapidly than the time required by longitudinal magnetization M_z to recover. Both processes run exponentially, but with different timing.

A certain time is required for the longitudinal magnetization to recover (represented by "T_1"). Transverse magnetization, however, disappears in a shorter time (represented by "T_2").

That is why there is no simple tilt back.

But what is more important: Beyond proton density, MRI has two additional contrast mechanisms that enhance diagnostic value.

These mechanisms are known as **relaxation**.



"Relaxation can be seen to act like a friction which counteracts the tilt produced by the alternating field."

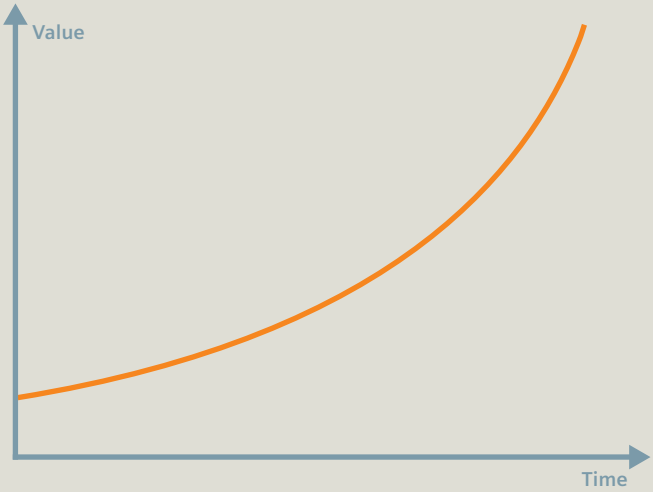
Felix Bloch
(1905–1983)

The Swiss-American physicist Felix Bloch (Nobel prize 1952) was the first researcher to establish the 'Bloch equations,' which describe the motion of the macroscopic magnetization after resonance. The solutions to these equations describe both relaxation processes independently, resulting in exponential recovery for longitudinal magnetization and exponential decay for transverse magnetization, with different time constants.

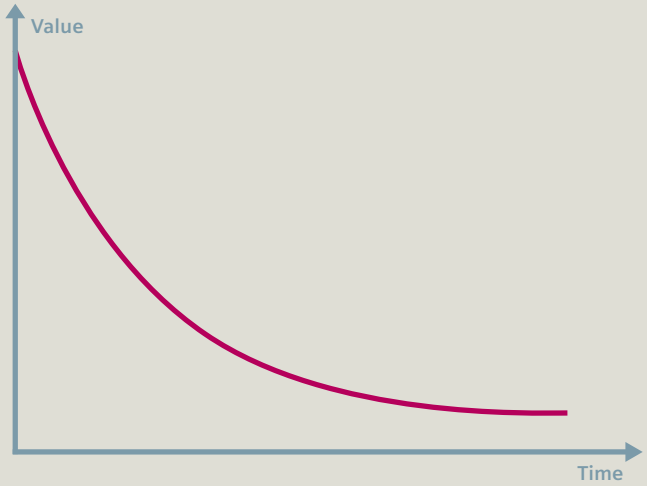
On exponential processes

Many natural as well as social processes can be expressed mathematically in a rather easy way: they are exponential. The increase in bacteria population, the reduction in radioactivity, compounded interest—all these are exponential processes. The same applies to spin recovery. This is reason enough for us to get involved.

Compounded interest is a good example of unchecked growth. Let us assume you have stocks or fixed rate funds in the value of $10,000 invested at a 10 % interest rate. After 10 years, your nest egg has grown to approximately $26,000, after 20 years to as much as $67,000, and after 50 years, you are a millionaire. By then, your savings have grown to $1.2 million.



Inflation is a good example of exponential decay. Let us say you start with a cash balance of $100,000 and a rate of inflation as high as 10 %, the value of your money would be down to approximately $34,000 in 10 years, hence drop to $12,000 in 20 years and be practically worthless after 50 years.



Relaxation made easy

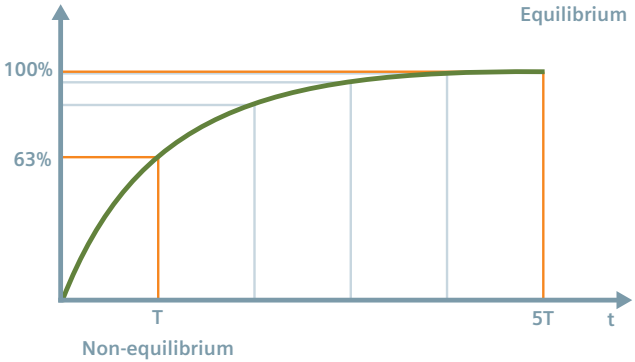
Timing is everything

The physical meaning of a time constant

An exponential process is expressed by its **time constant** T. After T, recovery of the physical quantity has reached 63 % of its remaining differential value, after 2T, it has reached 86 %, after 3T, it has reached 95 %, and after 5T, the process is nearly complete.

The strength of relaxation is a function of how far a physical quantity is from its point of equilibrium. The closer the system is to equilibrium, the weaker the relaxation. Thus, the process of returning to equilibrium slows down until a saturation value has been reached.

You can compare this to a rubber band: the higher the tension, the faster the rebound.

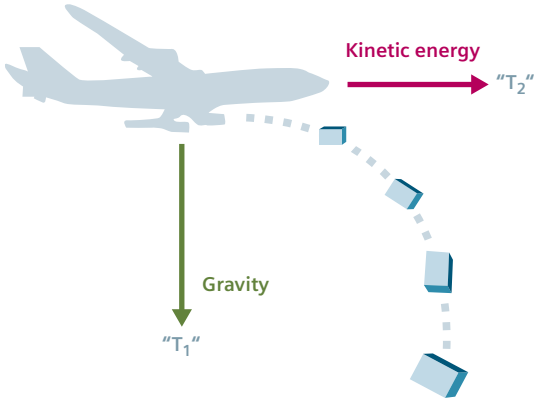


A falling box

We can compare the two different relaxation processes in MRI to a falling box. Imagine throwing the box from a high tower. It will travel with increasing speed toward the earth. The reason for this is the earth's gravity. So far so good.

If the box is thrown from an airplane, two simultaneous "forces" are at work: 1. gravity, 2. the kinetic energy in the direction of flight.

The actual movement of the box is the *superposition* of two motions performed independently of each other. While the box is falling toward the earth, coming closer and closer, it continues in the direction of flight but slows down in a parabola.

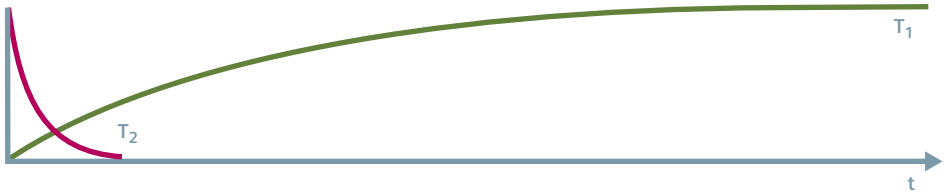


It is faster getting down the mountain than getting up

Let us summarize: while longitudinal magnetization is recovering, transverse magnetization is decaying. And as mentioned before, the decay of the transverse magnetization is usually *much faster* than recovery of the longitudinal magnetization.

The time constants are known as T_1 and T_2.

Normally, the T_2 constant is considerably shorter than the T_1 constant.



Longitudinal recovery

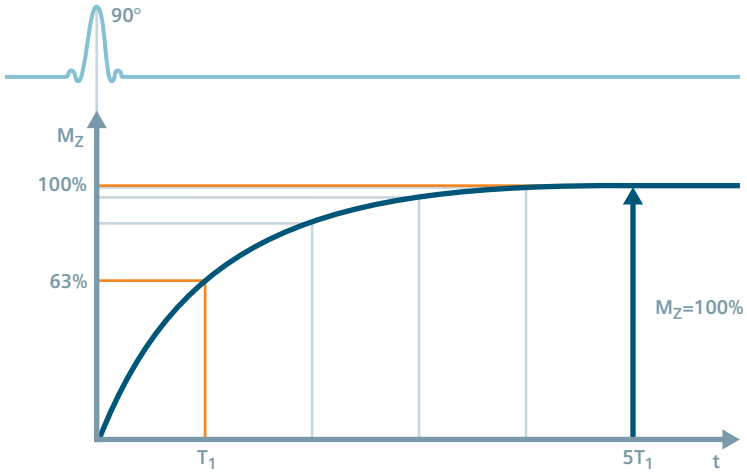
Moving back toward the initial maximum value with T_1

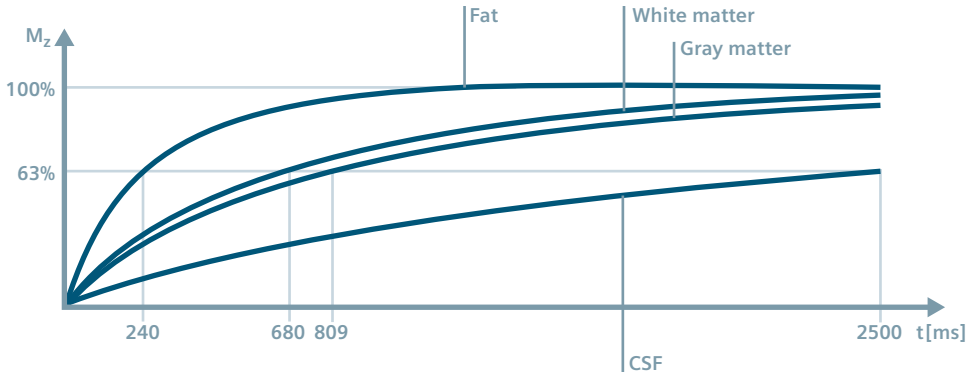
Return to equilibrium

The build-up and recovery of the longitudinal magnetization is an exponential process known as **longitudinal relaxation**. The time constant is known as T_1.

Please note: T_1 is *not* the time span of total magnetization recovery, but only the *time constant* for its exponential growth: After time interval T_1, the longitudinal magnetization M_z has recovered to approximately 63 % of its final value. After 5 T_1 times, recovery is practically complete.

What is important is that the T_1 constant depends on the tissue type.





The T_1 constant under the magnifying glass

Different types of tissue show different relaxation times. This is key to the sharp image contrast obtained with MR. As the table shows, the T_1 constant also *depends on the field* as well.

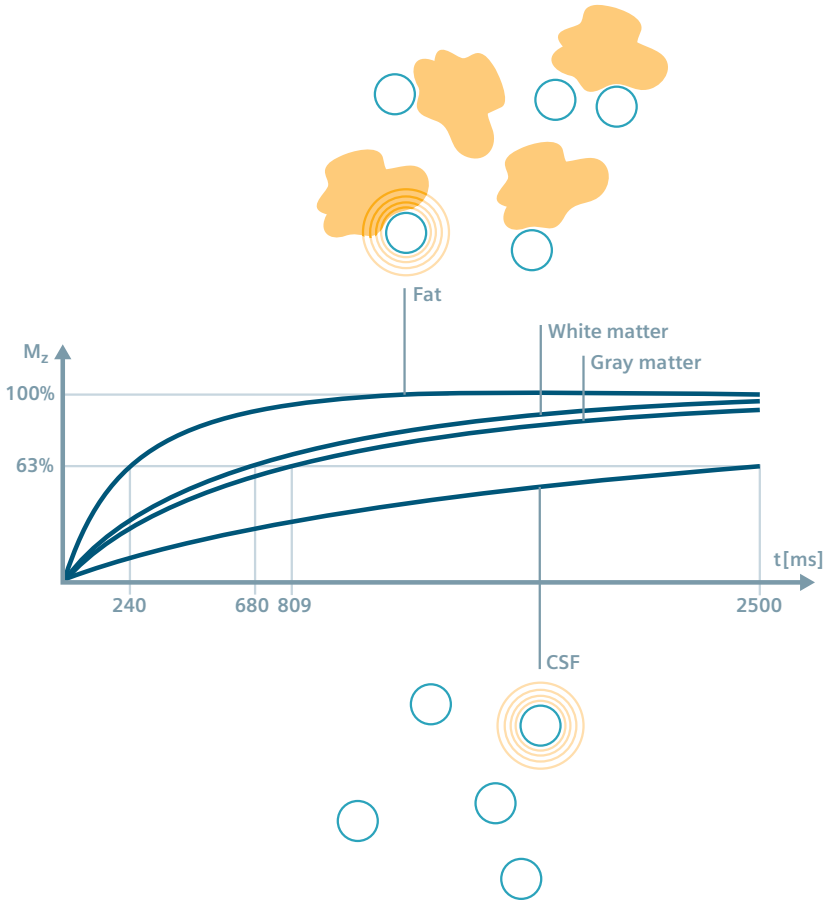
T_1 constants (in ms)

	0.2 tesla	1.0 tesla	1.5 tesla
Fat		240	
Muscle	370	730	863
White matter	388	680	783
Gray matter	492	809	917
CSF	1,400	2,500	3,000

Fat has a *short* T_1, water has a *long* T_1.

Causes of longitudinal recovery

Hydrogen protons in the body continuously "feel" local magnetic fields and their fluctuations caused by molecular motion. These minute field fluctuations add to the external field. The strongest effect is produced by field oscillations transverse to the main field that match the *Larmor frequency* of the protons. These "microresonances" cause the spins to randomly change their orientation with respect to the main field.



The T_1 process is frequently referred to as "spin-lattice relaxation" because in solid state magnetic resonance the molecular surroundings of nuclei are called a "lattice."

This process occurs after interference from an RF pulse and as early as during the build-up of the longitudinal magnetization, after the patient has been moved into the magnetic field.

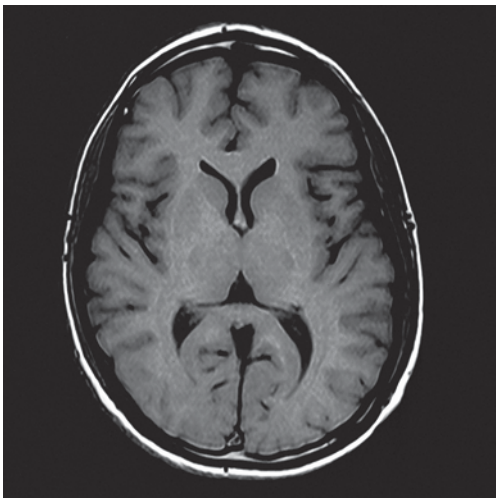
The environment of the protons frequently consists of larger molecules (lipids) and macromolecules (protein). Hydrogen protons inside a relatively slow-moving fat molecule as well as protons bound to proteins feel strong local field fluctuations: they quickly change their spin direction. This explains the relatively *short* T_1 relaxation of fatty tissue, for example.

In fluids, the molecular mobility of water is considerably faster than most field fluctuations. Resonances with oscillating magnetic fields are less frequent as well as weaker: the protons do not change their spin direction as quickly. This is why pure water and CSF show a relatively *long* T_1 relaxation.

Getting a taste for T_1 contrast

Since different types of tissue show different T_1 relaxation, this difference can be shown as MR image contrast. How this happens is explained in detail in a following chapter.

Diagnostic use in a nutshell: pathological tissue shows a different concentration of water than the surrounding tissue—and this means a different relaxation constant. The differ-ence in relaxation is visualized as contrast in the MR image.



With T_1 contrast, CSF appears dark in the MR image

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After resonance, the longitudinal magnetization fully recovers over a time that depends on the type of tissue and the field strength. This process is known as *longitudinal relaxation*.

The longitudinal relaxation follows an exponential course of growth characterized by the time constant T_1.

As the T_1 constant depends on the type of tissue, this characteristic can be employed for the MR image contrast.

The source of T_1 relaxation are local magnetic field fluctuations generated by molecular motion. Transverse field fluctuations in the range of the Larmor frequency show the strongest effect and cause the proton spins to change their direction randomly.

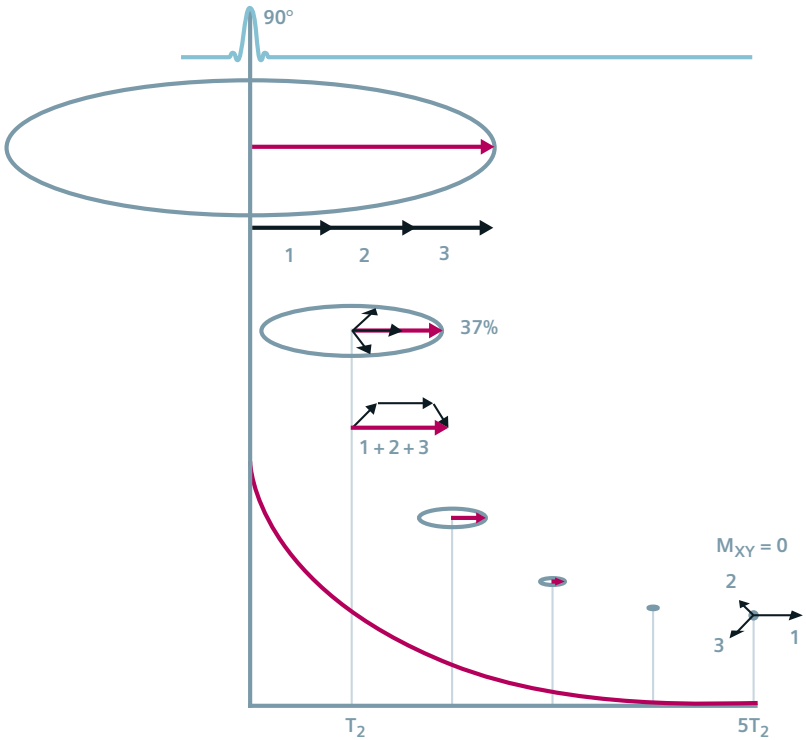
Transverse decay

The lifetime of the resonance phenomenon (T_2)

The signal must decay

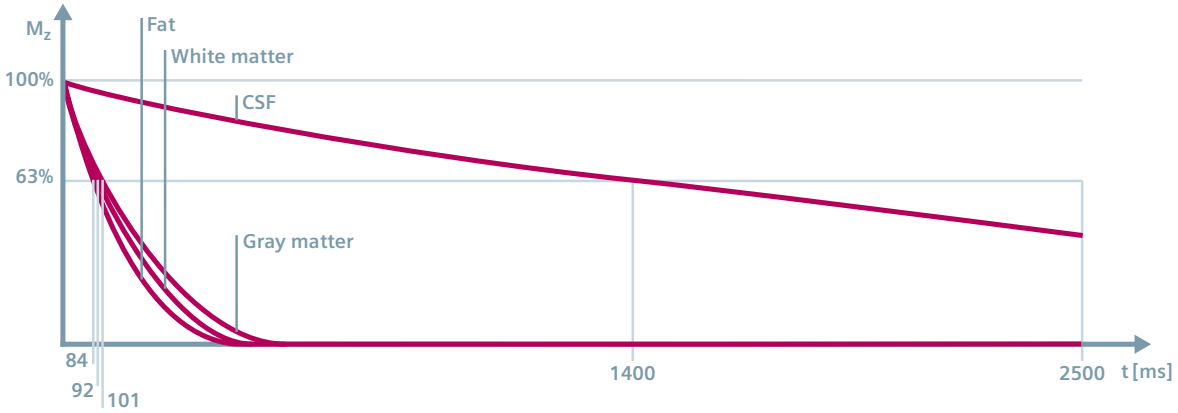
Directly after the RF pulse, the spins are what is called *phase-coherent*. They act like one large magnet that rotates in the xy-plane. That's why they can induce a signal in the receiver coil.

However, the rotating spins get out of phase again due to unavoidable molecular interactions, and the transverse magnetization begins to decay.



As the transverse magnetization decays, the induced MR signal deteriorates. This is what we call **transverse relaxation**. Its time constant is known as T_2. As we will see later on, this is only the maximum time for which magnetic resonance will persist. In reality, the MR signal (FID) decays at a more rapid rate.

After time interval T_2, the phase coherence of the spins has dropped to approximately 37 %. After 2 T_2, it drops to approximately 14 % and after 5 T_2, phase coherence has just about disappeared.



The T_2 constant under the magnifying glass

T_2 is also tissue-specific, but largely field-independent.

T_2 constants (in ms)

Fat	84
Muscle	47
White matter	92
Gray matter	101
CSF	1,400

Fat has a *short* T_2, water has a *long* T_2.

Reasons for the loss of coherence

There are two major causes of transverse relaxation.

1. As we have shown, fluctuating local fields are responsible for longitudinal relaxation. And it has also an effect to the transversal relaxation: when the proton spins change their direction randomly, their phase coherence is lost as well. This is the "T_1 contribution" to T_2.

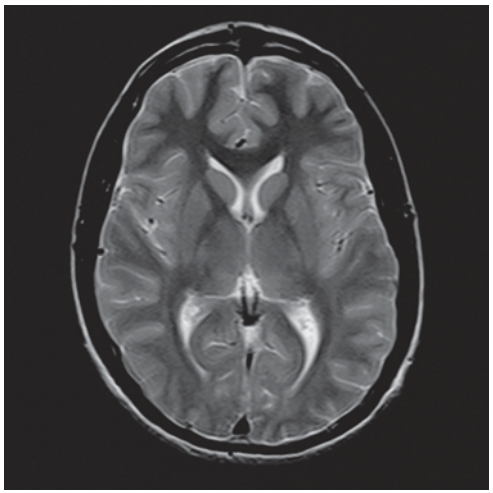
2. In addition, fluctuating magnetic fields in the direction of the external field make the local field vary by a small amount of approximately ± 1 millitesla. Accordingly, neighboring protons precess with slightly different frequencies of approximately 40 kHz around the normal Larmor frequency.

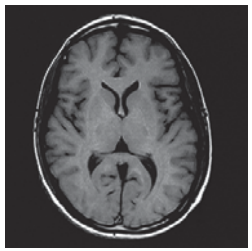
These slightly different precession frequencies are an additional reason for the loss of phase coherence: As a result, transverse magnetization decays before the longitudinal magnetization recovers.

Although the interaction between the spins is not the only source of transverse relaxation, the term "spin-spin relaxation" has made its entrance and is here to stay.

Getting a taste for T_2 contrast

Since different tissue types show different T_2 relaxation, these differences are shown as MR image contrast. A detailed explanation is provided in a following chapter.





T_2 contrast shows CSF bright in the MR image—opposite to T_1 contrast

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Immediately after the RF pulse, the rotating transverse magnetization starts to decay.

1. Spin precession gets out of phase until all spins are distributed uniformly within the transverse plane.

2. At the same time, longitudinal magnetization starts to recover.

Transverse relaxation follows an exponential decay curve characterized by the time constant T_2, which is a measure of the speed of spin dephasing.

The T_2 constant is also tissue-specific and contributes to the image contrast.

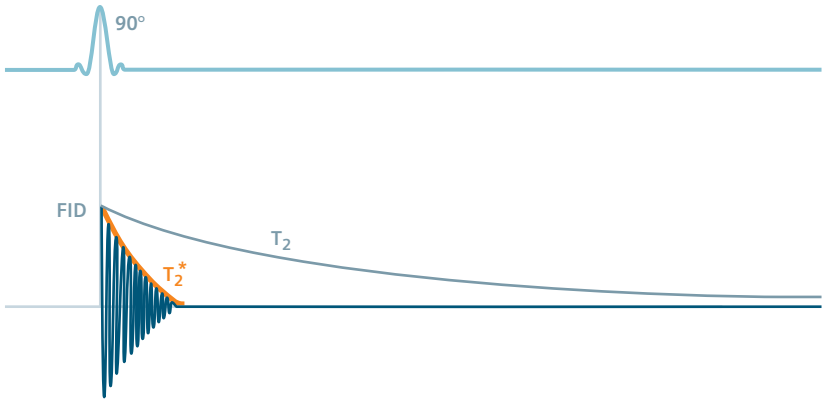
The spin echo

The magical moment: getting back the lost signal

The true decay of the FID

We might expect the MR signal (FID) to decay with the time constant T_2. However, the FID decays much more quickly, with a shorter *effective* time constant T_2^* (see figure).

The static magnetic field as felt by the spins is not the same everywhere, in fact it is somewhat **inhomogeneous**. Unlike the relaxation processes that lead to the T_2 decay, we are dealing here with purely static differences in the magnetic field that remain constant over time within a specific location. These are mainly spatial field variations caused by the patient's body as well as technical inhomogeneities of the main magnet.

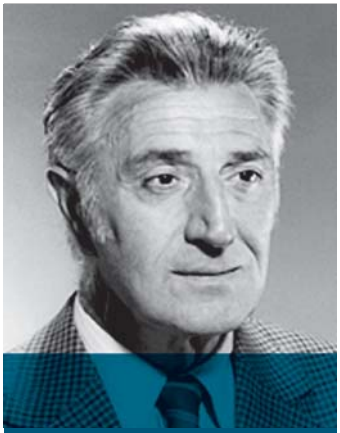


T_2^* decay

Static magnetic field inhomogeneities add to the fanning of the spins: they dephase more quickly than with T_2 relaxation. This is T_2^* decay.

It seems as if the phase coherence of the spins is irrevocably destroyed faster than T_2 relaxation "allows."

But this is not in fact the case, as demonstrated by the spin echo effect.



The U.S. physicist Erwin Hahn, born in 1926, discovered the spin echo in 1950: the recovery of the MR signal because T_2^* decay is reversible. Echoes can also be created by using inversion pulses different from 180 degrees, also called "Hahn echoes." Hahn originally used a pair of 90-degree pulses.

"The echo effect is brought about by subjecting the sample to two r-f pulses in succession. At time (TE) after ... the second pulse the echo signal appears."

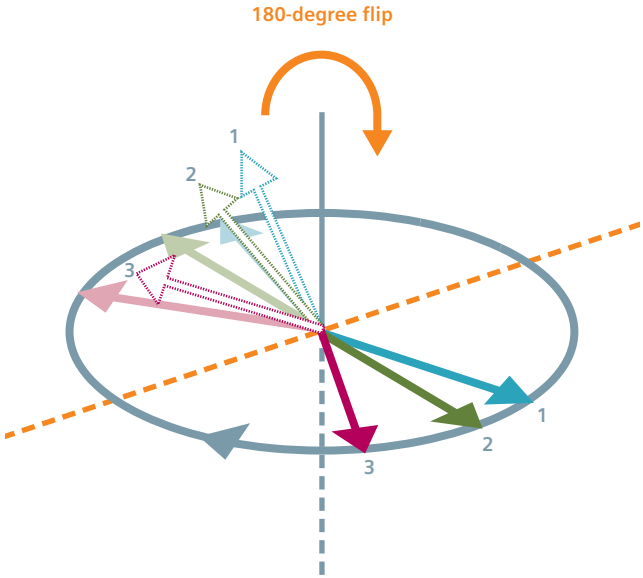
Erwin L. Hahn

Spins flipped like an omelet

Since static magnetic field differences remain constant with respect to space and time, we can cancel out their effect with a "magic trick."

We give an "about-face" command via a 180-degree pulse. The 180-degree pulse apparently flips the spins just like an omelet: the order of the spins is reversed, the direction of precession remains, of course.

The faster spins (1), now placed behind, catch up with the slower (3) precessing spins.



Erwin Hahn originally explained the echo effect from the analogy of a team of runners with different running speeds. During the race, the group of runners fan out more and more, seemingly in a random manner. They are asked to turn around by 180 degrees and then continue running in the opposite direction. In this way, they will return together precisely at the starting line. We are experiencing an "echo" of the start.

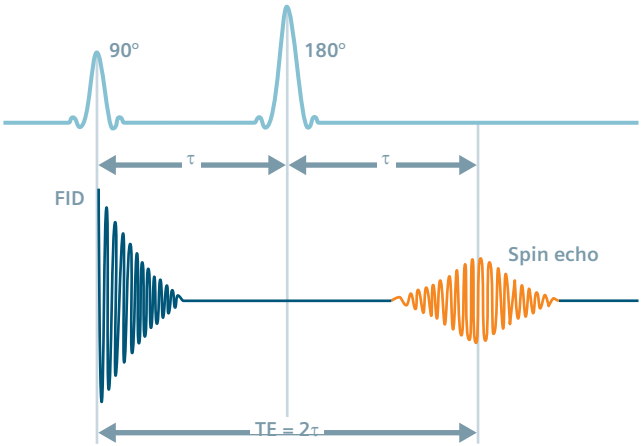
But there is a catch: the spins don't turn around by a 180-degree pulse. This would only happen by reversing the main magnetic field. In fact, the spins always precess in the *same* direction, for example clockwise. That is why we prefer the "flipped omelet" analogy.

Here comes the echo

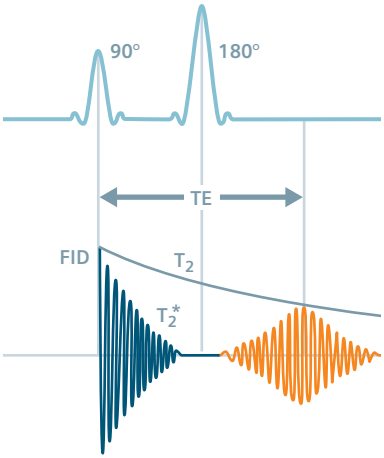
This is the effect of the 180-degree pulse: the out-of-phase spins go back into phase, and a new MR signal is generated—the **spin echo**.

The 180-degree pulse is switched after the 90-degree pulse after the run time τ. The spin-echo signal initially increases and reaches its maximum after double the run time (2τ). This time span is known as the **echo time** (TE). The spin echo will decrease after this time.

Since the FID begins to decay immediately after the 90-degree pulse, it is difficult to measure its strength. For this reason, echoes are the preferred signals for imaging.



Please note: The spin-echo signal decays with T_2^*, its strength (amplitude, maximum), however, depends on the T_2 relaxation curve.



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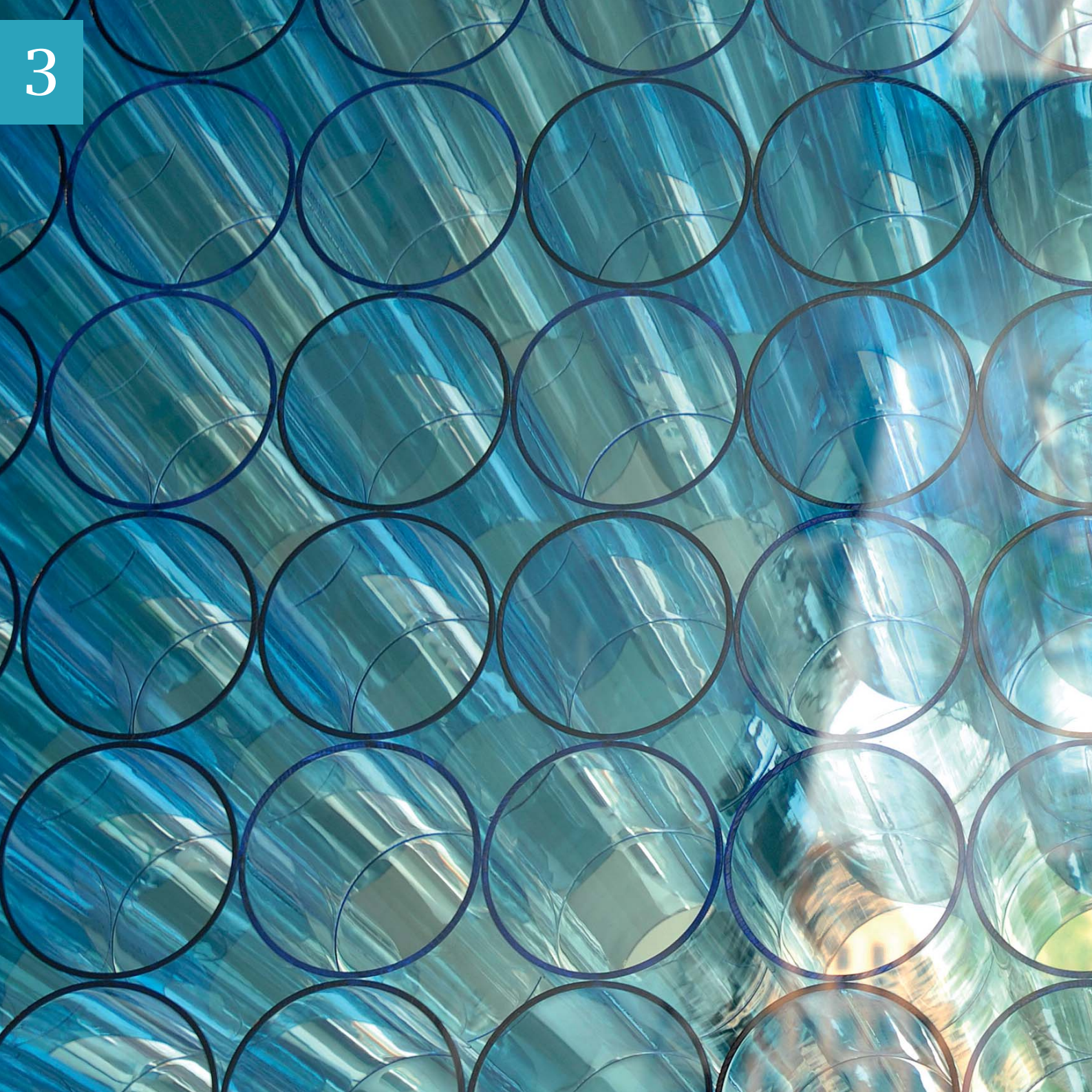
The MR signal (FID) decays according to the very short time constant T_2^*, smaller than T_2. The cause of this fast decay are static magnetic field inhomogeneities. They force spins to go out of phase more quickly.

We can recover the MR signal with a second pulse, usually a 180-degree pulse. This MR signal is known as the spin echo.

Rule of thumb:

$T_2^* < T_2 < T_1$



From the Signal to the Image

How do we generate an image from an MR signal that shows spatial structures as different gray values?

Slicing by gradients

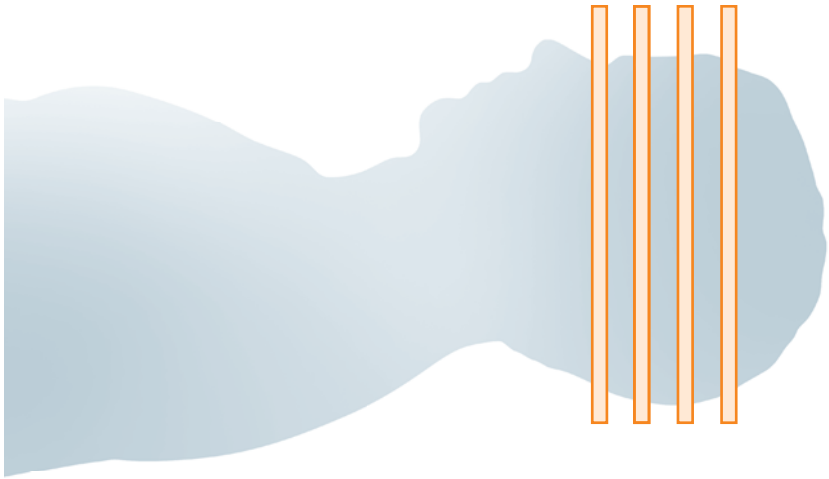
Creating “zip codes” for locations in the body

The imaging question

In tomographic imaging, we create slice images at specific positions of the human body.

Remember that, in the magnetic field of the MR scanner, a spatial distribution of longitudinal magnetization is built up in the body. If, for example, we simply excite all the spins within the head with one RF pulse, the averaged transverse magnetization in the head would generate the MR signal. We would not obtain any spatial resolution.

The question remains: how can we distinguish between MR signals from different locations in the body?

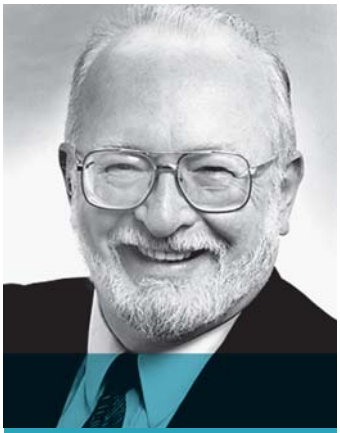


The trick with the gradients

As the magnetic field of an MR scanner is not perfectly homogeneous, spins will precess at different frequencies. But only those spins that precess at the correct Larmor frequency will resonate with the RF pulse. In order to make the field more homogeneous, a technical process known as 'shimming' is applied.

The revolutionary idea, which started MRI, was to 'unshim' the magnetic field in a controlled way: by switching **gradients**.

As a result, spins at different locations in the body will precess at different frequen-cies, which is intended.



"The distribution of magnetic nuclei such as protons ... may be obtained by imposing magnetic field gradients ... and measuring the intensities as functions of the applied magnetic field."

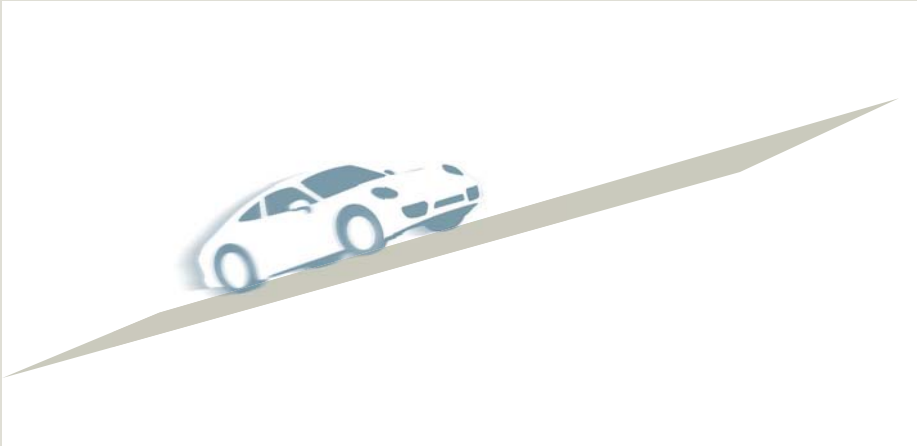
Paul C. Lauterbur
(1929—2007)

Professor Lauterbur's insight: why not make the proton spins from different locations precess at different frequencies on purpose? Instead of placing the body in a uniform magnetic field, we let the field vary from one location to another. We can use the different frequencies that make up the mixed MR signal as a "zip code" for corresponding anatomical locations. Paul Lauterbur applied this idea for the spatial encoding of voxels within a slice. The idea of slice selection by gradient fields was introduced by Sir Peter Mansfield.

On gradients

A gradient is an incline comparable with the incline of a road. From a mathematical point of view, a gradient defines the strength and direction of a magnitude changing in space.

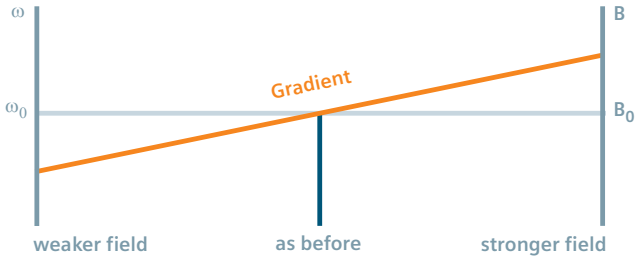
Translated into MR technology this means: a **magnetic field gradient** is a controlled change in the magnetic field strength in a certain direction, that is, a linear increase or decrease.



Locally changing the field strength

In a homogeneous magnetic field, the field strength is the same everywhere by defi-nition (B_0). For this reason, all proton spins show the same precession frequency ω_0 proportional to the field strength. The magnetic resonance is the same everywhere.

When a magnetic field gradient is applied, the field shows a linear increase. The precession of the proton spins in this direction varies accordingly. They precess more slowly in a weaker field and more quickly in a stronger field. In sum they show *different* resonance frequencies.



Defining the slice position

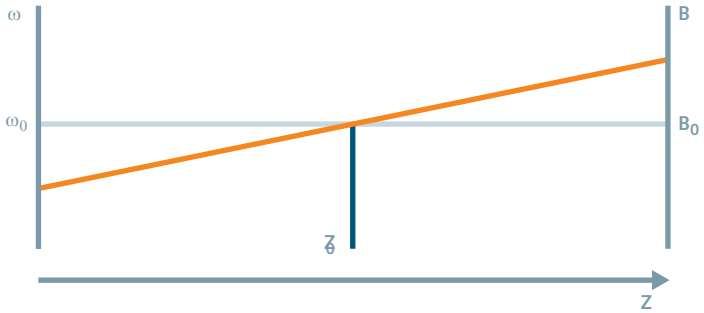
Slice positioning and slice thickness

The slice-selection gradient

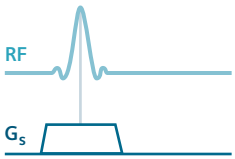
For slice selection, a gradient is switched in the z-direction, for example, *simultaneous-ly* with the RF pulse. This gradient is called the **slice-selection gradient** (G_S).

Now the field has its original strength B_0 at one location only z_0. If the RF pulse had only one single frequency ω_0, it would only excite the spins at the resonance location z_0. This is the selected **slice position**.

However, this does not really suffice. We would get a "slice," but without thickness. The slice would be paper thin and the signal too weak, because only a few protons would be excited in this thin area. What we need is a certain resolution in the z-direction. And this is what we call **slice thickness**.



The strength of the homogeneous static magnetic field is B_0. The associated Larmor frequency of the proton spins is ω_0.



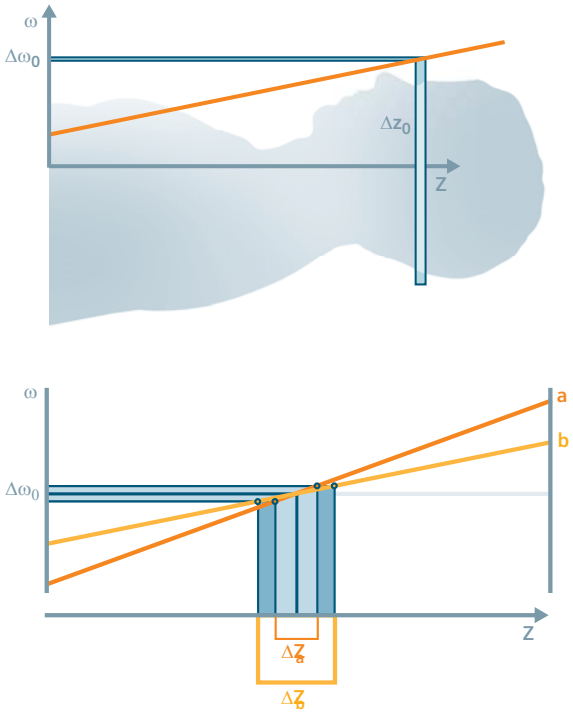
How do we select the slice thickness?

The RF pulse has a certain **bandwidth** of neighboring frequencies about its center frequency ω_0. So it can excite the desired spatial area of the slice thickness (Δz_0).

As an alternative: the slice thickness can be modified by keeping the RF pulse bandwidth constant while changing the gradient strength. A steeper gradient ramp (a) excites a thinner slice (Δz_a), a shallower gradient (b) excites a thicker slice (Δz_b).

Whatever technique is used: a **slice** is a *defined resonance area* of the proton spins. Outside the slice, the spins are not excited by the RF pulse. Transverse magnetization (and therefore an MR signal) is generated only within the selected slice.

A slice is defined by switching a magnetic field gradient.

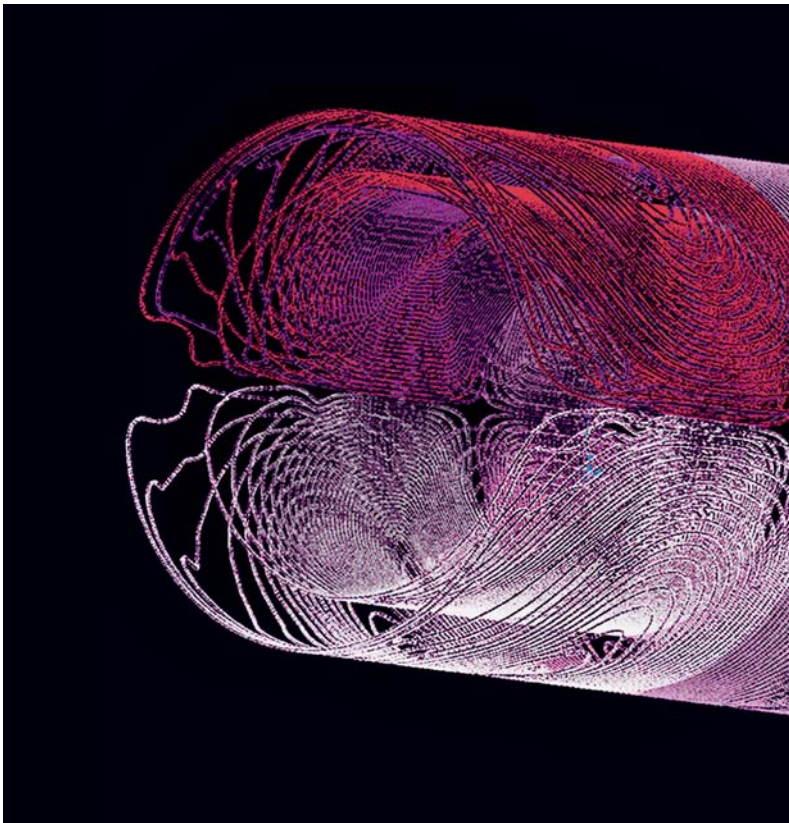


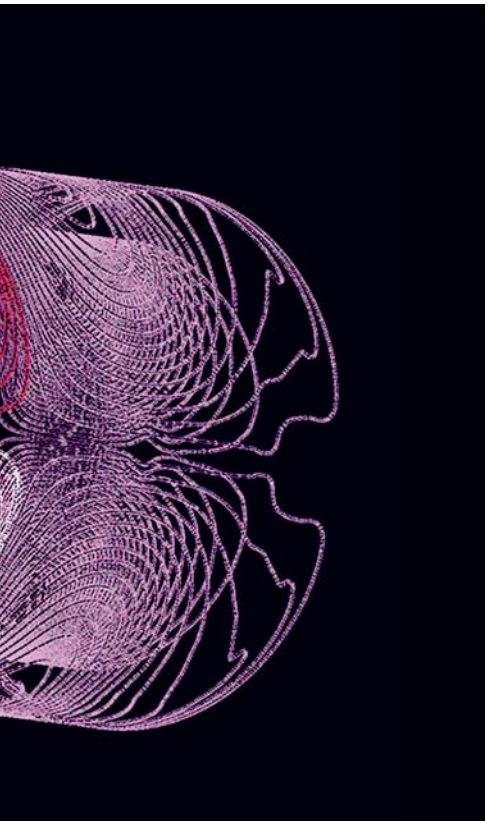
This is how gradients are generated

Gradient coils

The MR system has three gradient coil arrangements for all three spatial directions (x, y, and z), situated around the magnet bore. The gradient coils do not generate a permanent magnetic field, but are switched on briefly and multiple times during the examination.

The gradient coils are operated via special power supplies, known as gradient amplifiers.





Gradient coil design for an MRI scanner

Gradient performance

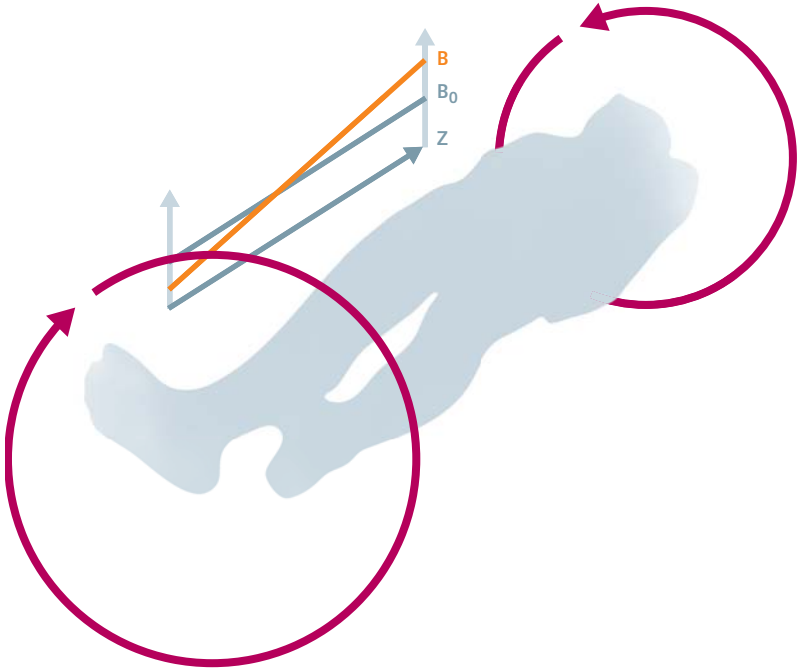
The performance of a gradient system is characterized by the minimum rise time required to obtain the maximum amplitude (=gradient strength). The **rise rate** is calculated from these two parameters. These characteristic data are also known as the **slew rate** (SR) and allow for a quick comparison of the performance of gradient systems.

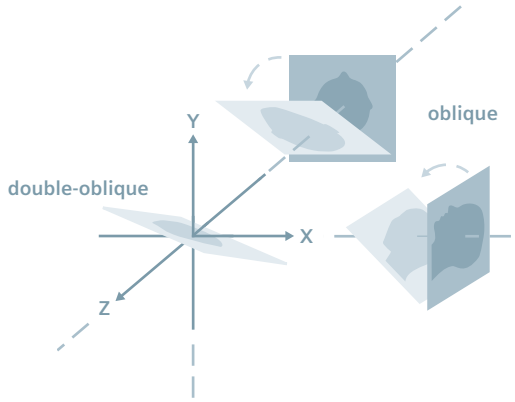
How do gradient coils work?

As a principle of physics, a magnetic field is created as soon as an electric current flows through a conductor or a coil. When the direction of current is reversed, the orientation of the magnetic field changes as well.

In MR, **gradient coils** in pairs are operated in a specific direction, for example, the z-direction, with the *same* current strength but *opposite* polarity.

One coil *increases* the static magnetic field, the opposing coil *reduces* it. This means that the magnetic field with its original strength B_0 changes like the incline of a road.





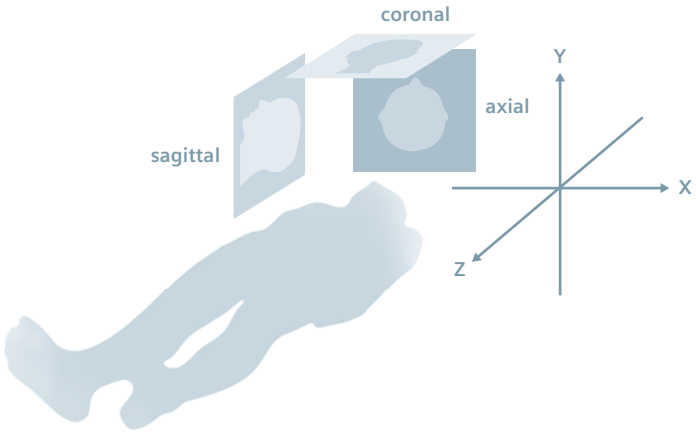
Free slice positioning

The huge advantage of gradients in MRI: they allow us to position arbitrary slice planes.

For sagittal slice positioning, we need to switch the *x-gradient*, for coronal slice positioning, the *y-gradient*.

To obtain **oblique** slices, we have to switch two or three gradients simultaneously. Their effect is superimposed.

A single oblique slice is generated by two gradients, for example, in the z and y-direction. For a double-oblique slice, the system switches all three gradients simultaneously.



The gradient system: safety aspects

Noise

As with a loudspeaker, strong mechanical forces are generated by the gradient coils, resulting in knocking noises during the examination. These noises are attenuated by using suitable measures, for example, earplugs or special sequences, for example, whisper sequences or "quiet sequences."

Pacemakers

Pacemakers are critical with respect to gradient fields. The control as well as the programming of pacemakers may be adversely affected by high-speed switching gradient pulses.

Physiological stimulation

At certain thresholds for the rise time and amplitude of the gradient fields, the induced voltages may be large enough to cause peripheral nerve stimulation. The muscle fibers contract involuntarily. This is not hazardous to the patient's health but may, however, be uncomfortable for the patient.

In the safety standards for MR systems, the maximum field changes are defined as a function of the switching time. Normally, these threshold values are not exceeded by the imaging methods used today. However, with some sequences, which use extremely fast gradient switching, for example, EPI sequences, these thresholds can be exceeded. But for safety purposes, gradient pulses are limited in routine applications.

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By switching gradients, we can make spins precess at different frequencies in a controlled manner.

The slice-selection gradient allows us to generate a slice, a spatial region where proton spins will resonate. Outside the slice, the spins are not affected by the RF pulse.

By switching two or three gradients simultaneously, we can position arbitrary slice planes.

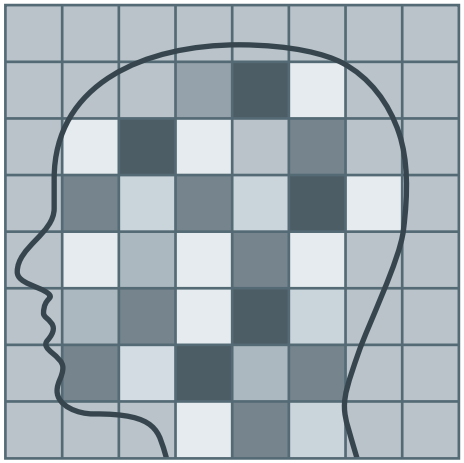
The matrix: resolution

Encoding and decoding the "zip codes" that make up the image

Defining the image matrix

We have defined the slice position for an image. Now we want to define the image matrix and its resolution. What interests us: how do we get an image from a slice?

Let us assume we want to generate a tomogram in a 256×256 **matrix size**. In this case, each row and column has to be divided into 256 locations. We use the MR signal to create an image with 256 different values, that is 65,536 voxels. This is spatial resolution.



Following Monsieur Fourier

What is important here is: the image is *not* directly generated from the measurement procedure. Instead, by switching additional gradients and using some signal processing, raw data are generated first, in our example 65,536 data points. The final image is computed from the raw data.

These processes include some signal encodings and mathematical transforms.

Let us follow these processes step by step.

The French mathematician and physicist Jean-Baptiste Joseph Fourier developed his series of sine and cosine curves while investigating the propagation of heat. This mathematical tool helps to analyze and construct a multitude of phenomena, including the MR signal.

"A general problem consists in developing any function whatsoever in an infinite series of sines and cosines..."

Joseph Fourier
(1768–1830)

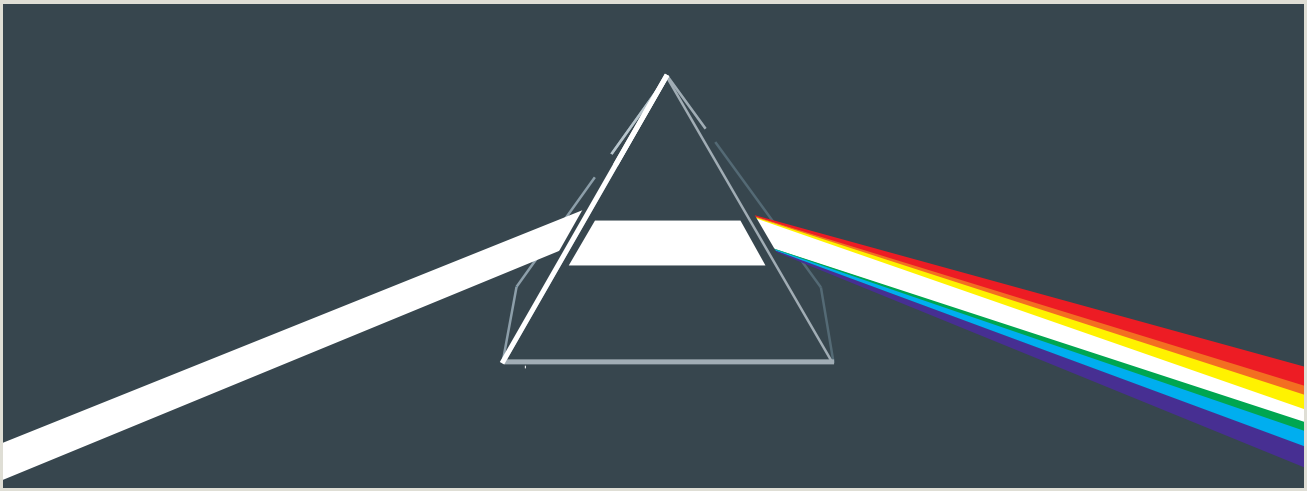
On prisms and spectra

Almost all natural and technically-generated signals comprise a mixture of oscillations of different frequencies.

White light is a mixture of light consisting of different wave lengths or frequencies. A prism helps us to differentiate the various frequencies contained in white light—in this case the colors of a rainbow.

The same applies to hearing: Most sounds consist of a multitude of acoustical waves and pitches, which our sense of hearing can differentiate, for instance, in a piece of music.

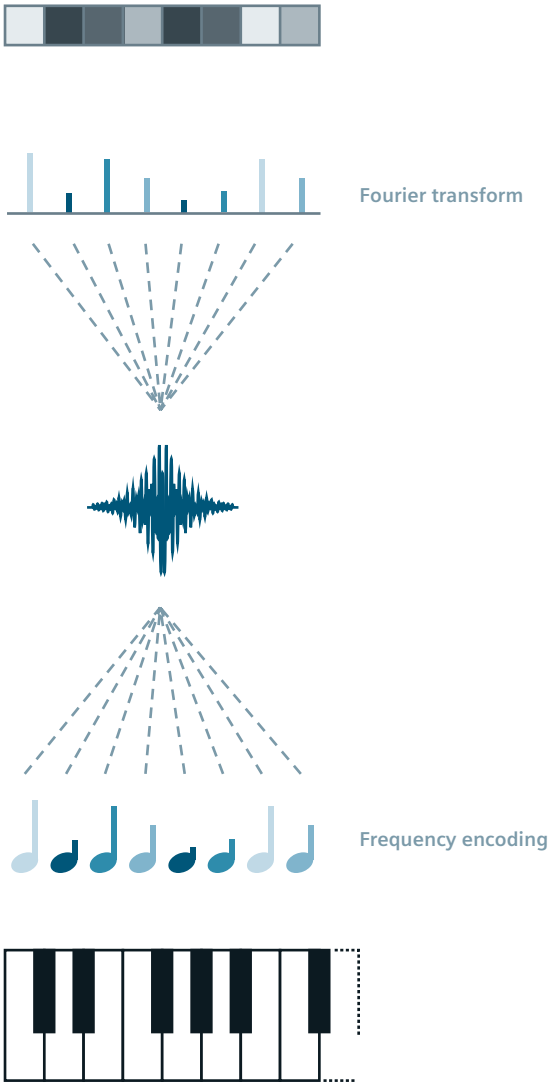
This spread across frequencies is called a spectrum. The mathematical method for investigating such a sprectrum is known as Fourier analysis.



We can build a given structure from the "building blocks" of sine curves. The more curves we use, the finer the results. The curve showing the profile is the result of 32 superimposed sine curves: Fourier synthesis.

A stripe image

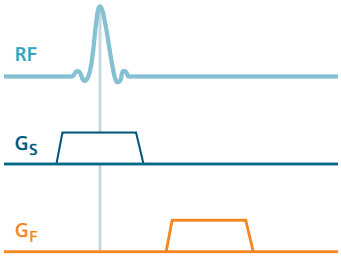
"Zip code" by frequencies



The piano analogy

Imagine a piano keyboard, the white keys representing eight different notes, i.e. frequencies, of an octave. Each key is simultaneously hit with a different strength, resulting in a mixture of notes. Can we, just by listening, know which notes make up this audible experience, and what the respective amplitudes are?

The MR scanner can. The MR signal is actually made to be a *mixture* of the signals of all excited spins along, say, the x-axis. At a resolution of 256 voxels, an echo includes not only eight but 256 "notes" of different frequencies.



This is frequency encoding

During the measurement of the echo, we switch a gradient in the x-direction. What is going to happen?

As you already know from gradients, the spin ensembles of the individual voxels precess along the x-axis at an increasing frequency. This is known as **frequency encoding**. The associated gradient is known as the **frequency-encoding gradient** (G_F).

The echo is a mixture of signals. It contains many sine and cosine waves. By knowing the gradient strength, we know the frequencies left and right and at all positions along the gradient.

The **Fourier transform** allows us to determine the signal contribution of each frequency component (as shown in the graphic by the height of the peaks). The individual frequencies are reallocated to their location of origin along the x-axis. The individual signal strength obtained determines the gray value of the allocated pixel.

For the definition of a line of voxels, the frequency-encoding gradient is switched.

Piling up stripes for a complete picture

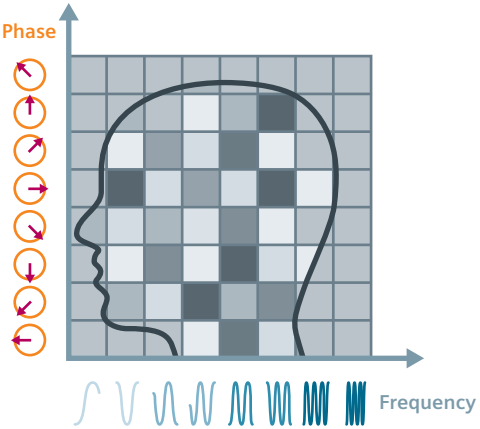
"Zip code" by phases

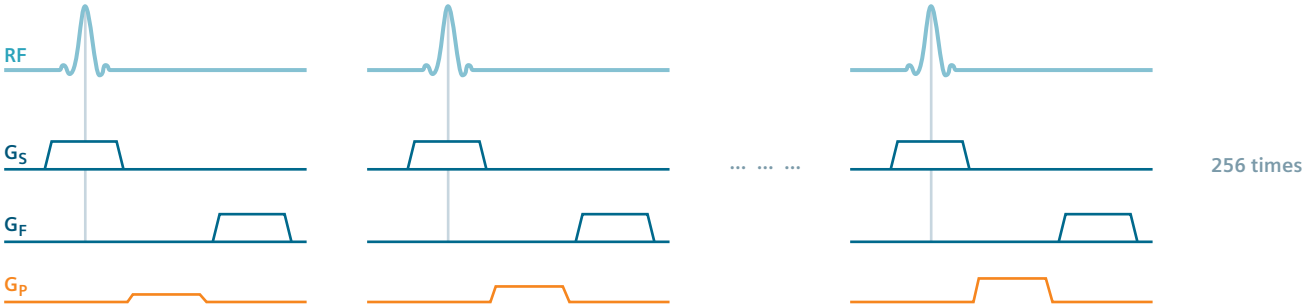
This is phase encoding

We cannot use the same frequency-encoding trick in the y-direction to encode a two-dimensional image. Why not? In this case, two different voxels could have the same frequency and thus could not be differentiated. Obviously we have to choose a different method.

During the time *between* the RF pulse and the echo, a gradient is switched briefly in the y-direction. As a result, the spins precess at different speeds for a short time. After the gradient is switched off, the spins along the y-axis show different phase shifts directly proportional to their locations.

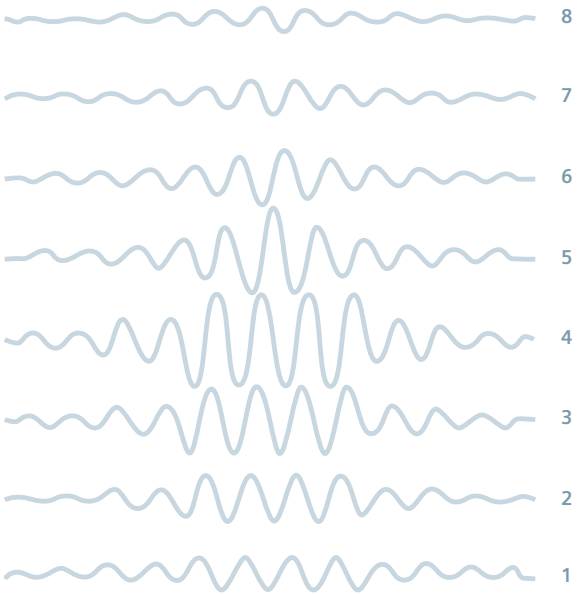
This process is called **phase encoding**. The associated gradient is the **phase-encoding gradient** (G_P).





We can use the Fourier transform to filter out these phase shifts as well. There is one slight drawback, however, it takes more time. We have to generate 256 MR signals with different phase encodings for 256 different locations. In other words, this means 256 **phase-encoding steps**. This is why the pulse sequence has to be repeated 256 times for a 256 × 256 matrix.

Thus, a **signal matrix** is filled line by line with the echoes (shortened to eight in the graphic). The lines represent frequencies, the columns represent phases.



For the definition of multiple lines of voxels, the phase-encoding gradient is switched.

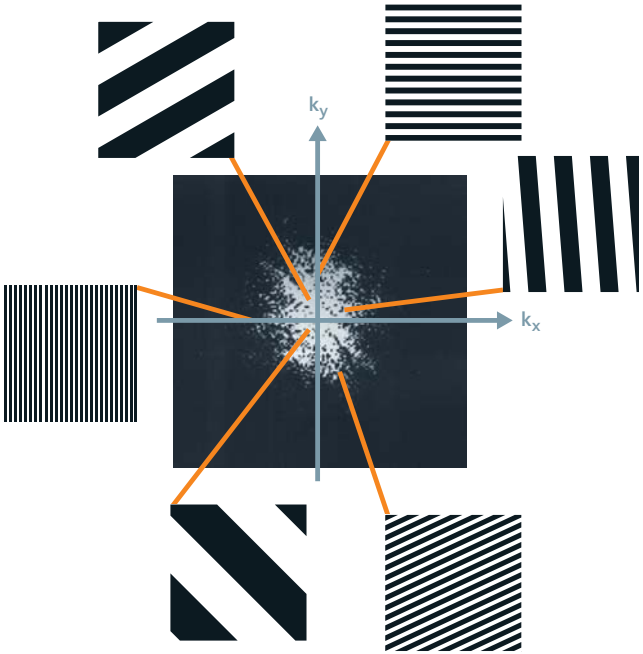
Transforming to and from k-space

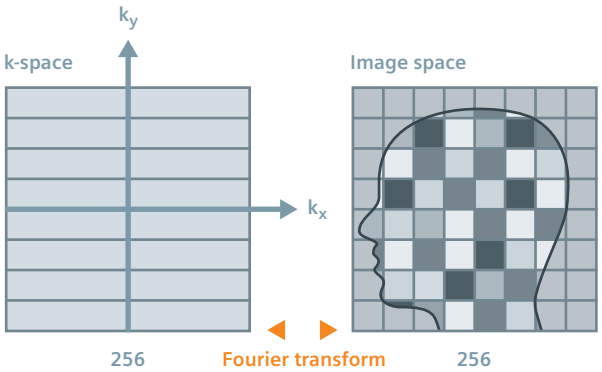
A matrix of stripe patterns

The signal matrix is preprocessed, resulting in the raw-data matrix, also known as **k-space** (a notion adopted from wave physics). The axes (k_x and k_y) of the raw-data matrix designate so-called "spatial frequencies". What do you think they are?

Just as temporal oscillations combine waves of different frequencies (sines and cosines), so an image can be composed from different spatial waves or stripe patterns.

The raw data simply determine whether and how intensively a certain stripe pattern contributes to the image. A rough stripe pattern (close to the center) shows low spatial frequency. A fine stripe pattern (far from the center) shows high spatial frequency.

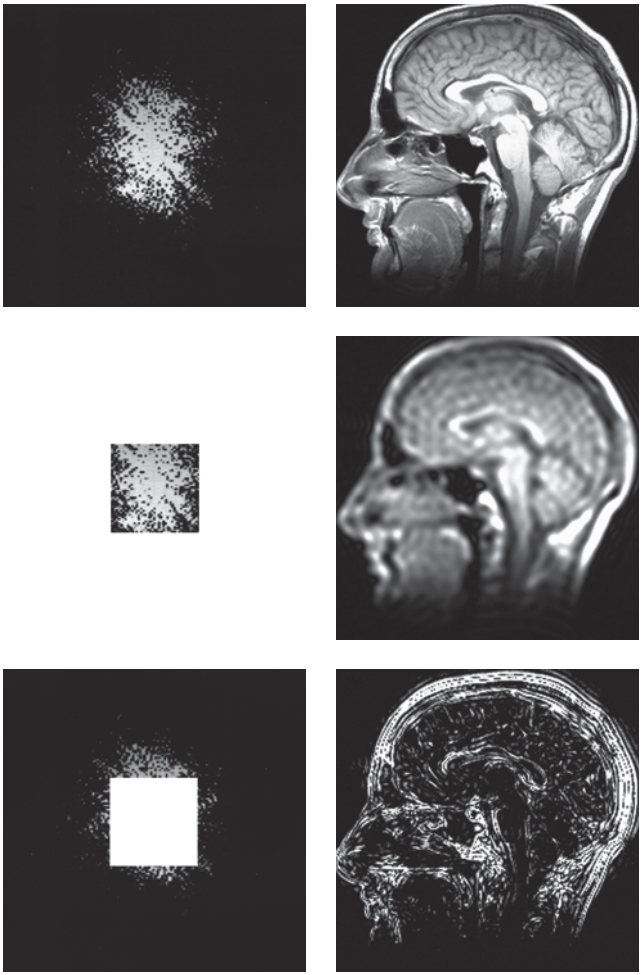




An example will help to illustrate our case: by simply superimposing the horizontal and vertical stripe pattern shown, we generate a complex gray value pattern. You can see that a weighted superimposition of stripe patterns will result in a more complex image.

This is exactly what, as the last step in image formation, a **two-dimensional Fourier transform** does. It uses the raw-data values in the k-space to calculate the gray value distribution in the image, that is, the weighting of the many stripes. Subsequently, the associated gray value is assigned to each pixel. And there you have your image.

The MR image is created by a 2D Fourier transform from the raw-data matrix.



Raw data versus image data

This is interesting: a data point in the raw-data matrix (k-space) does *not* directly correspond to a pixel in the image. Each part of the raw-data matrix contains information about the whole image—comparable to a hologram.

The center raw data determine the rough structure as well as the image contrast.

The outer portions of the raw-data matrix provide information about margins, edge transitions, contours in the image—in short they show finer structures and in the final analysis determine the resolution.

Center raw data : structure and contrast
Outer raw data : resolution

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The MR imaging technique does not create the image directly. To localize the individual voxels, phase encoding and frequency encoding is used. For this, gradients are switched.

A signal matrix is filled with the resulting echoes and converted to a raw-data matrix, known as k-space.

The MR image is computed from the raw data by means of a final two-dimensional Fourier transform.

Introducing the pulse sequence

The complete picture

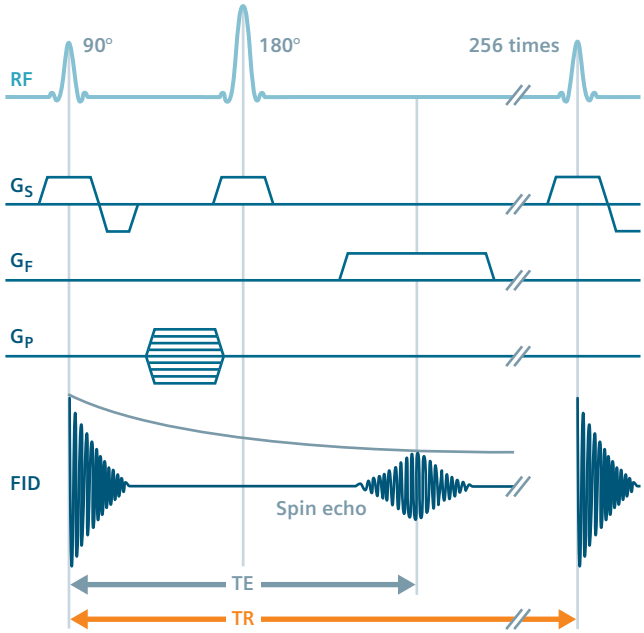
Understanding pulse diagrams

As you will remember, a spin echo is created by a 90-degree pulse, which generates the FID, followed by a 180-degree pulse, which generates the spin echo in echo time TE. This is a simple **pulse sequence**.

The pulse sequence is repeated with **repetition time** TR until the raw-data matrix is filled with echoes. The number of phase-encoding steps (that is, raw data lines) corresponds to the number of repetitions of the sequence. The scan time is determined to a large degree by the resolution of the image in the phase-encoding direction.

$Scan\ time = N_P \times TR$

(N_P: number of phase-encoding steps)

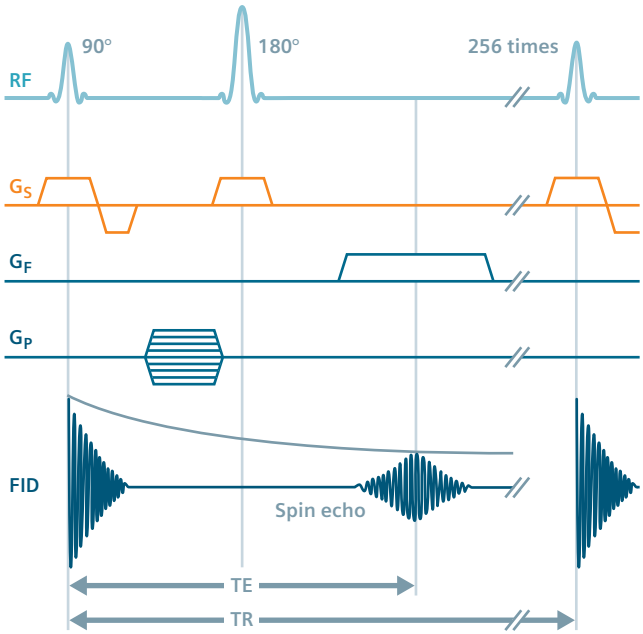


Slice encoding

The **slice-selection gradient** G_S is switched simultaneously with the 90-degree pulse.

What does the additional bar pointing downward G_S mean? The gradient dephases the spin phases along the slice thickness. We need to compensate for this with a gradient of opposite polarity and half the duration (rephasing gradient).

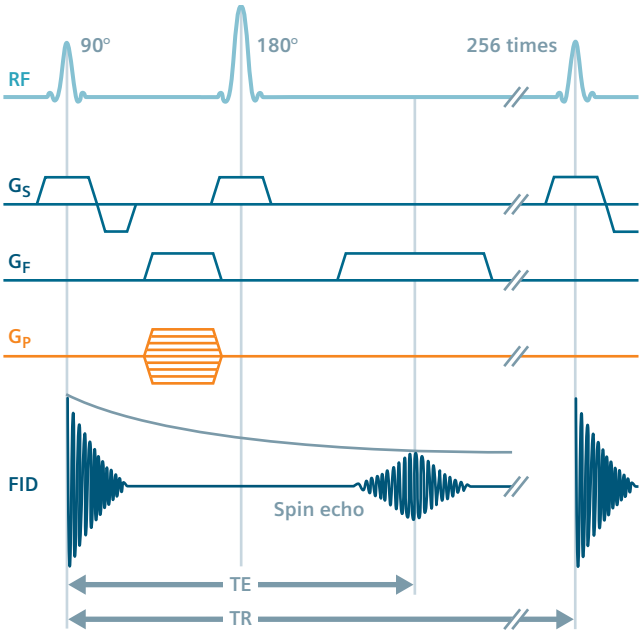
During the 180-degree pulse, the slice-selection gradient is switched again so that the 180-degree pulse only affects the spins of the previously excited slice.



Phase encoding

As you will remember, a phase-encoding gradient superimposes a different phase on spins from different locations, for example along the y-direction. The **phase-encoding gradient** G_P is switched on briefly between slice selection and spin echo.
For a matrix consisting of 256 columns and 256 rows, gradient switching of the spin-echo sequence is repeated 256 times with repetition time TR—with the phase-encoding gradient increasing step by step.

The phase-encoding steps in the pulse diagrams are frequently represented by a multitude of horizontal lines in the bar. These represent the different gradient step amplitudes—positive or negative.

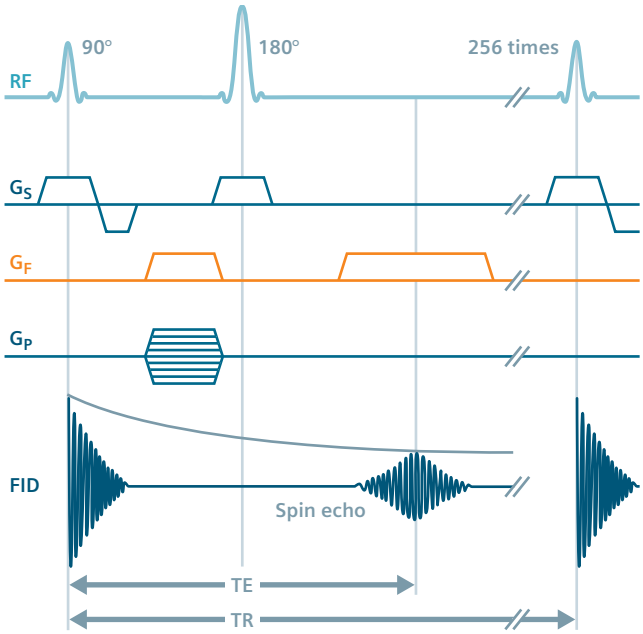


Frequency encoding

During the spin echo, the frequency-encoding gradient G_F is active. Since the spin echo is read out during this time, this gradient is also known as the **readout gradient**.

If we only applied the readout gradient, the spins precessing in the direction of frequency encoding would begin to dephase. During echo time TE, the spins would be fully dephased, leaving us without a spin echo. However, we can circumvent this problem by applying a dephasing gradient before the 180-degree pulse. This makes the readout gradient rephase the spins in a way that causes the spins in the center of the readout interval to be in phase again at the time of the maximum spin echo.

When we switch on the dephasing gradient before the 180-degree pulse, the gradient has the same polarity as the readout gradient. The 180-degree pulse reverses the phase of the spins.



The principle of MR imaging

By switching gradients, we obtain the signal mixture for a slice image in two steps:

- We only excite spins within a certain slice (slice selection).
- Subsequently, we acquire a 2D scan matrix via frequency and phase encoding in the slice.

With the help of two-dimensional Fourier transform, the MR system reconstructs the MR image from the raw data measured.

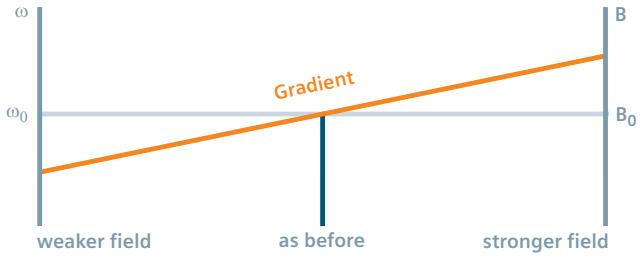
Meet the gradient echo

Spin echo's counterpart

Gradient pulses again

Let us recap the effect of a magnetic field gradient. When a gradient pulse is switched in a specific direction, spins will precess in this direction with linearly decreasing and increasing frequencies.

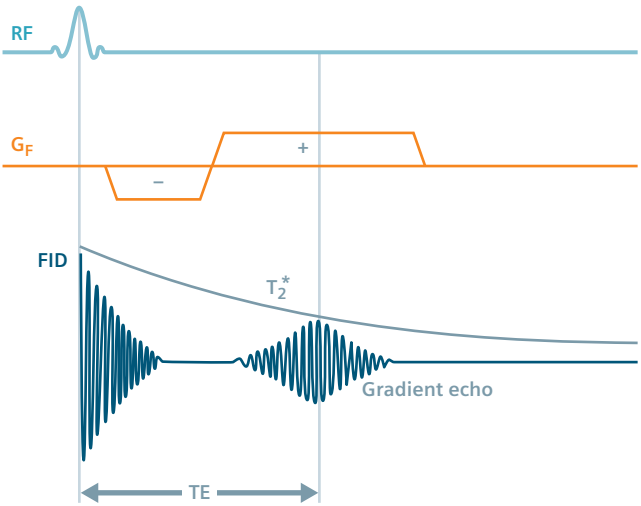
As a result, by using a gradient pulse we can destroy the free induction decay faster than it would if undisturbed.



A simply different echo

A gradient pulse (–) directly after the RF pulse artificially dephases the spin frequencies. Since they are now precessing at different speeds (faster on one side, slower on the opposite side), they lose their phase more quickly, that is, they are being **dephased**. The FID is eliminated considerably faster than it would under normal conditions.

When the polarity of the gradient is reversed (+), the precessing spins will be in phase again, that is, they are **rephased**. We measure an echo during the rephasing of the FID. Since this echo is generated by a gradient, it is called the **gradient echo**.



Only a little time for the echo time

The echo time TE is typically shorter for a gradient-echo sequence than for a spin-echo sequence. Why is that?

The 180-degree pulse is omitted in gradient-echo technology. This means that we do *not* cancel the static T_2^* dephasing mechanism as we do in spin-echo technology. Instead, we use gradient pulses to quickly destroy the FID and build it back up again, all within the T_2^* decay.

The echo time for a gradient echo has to fit into the T_2^* time. This is why the gradient-echo technique is faster than the spin-echo technique.

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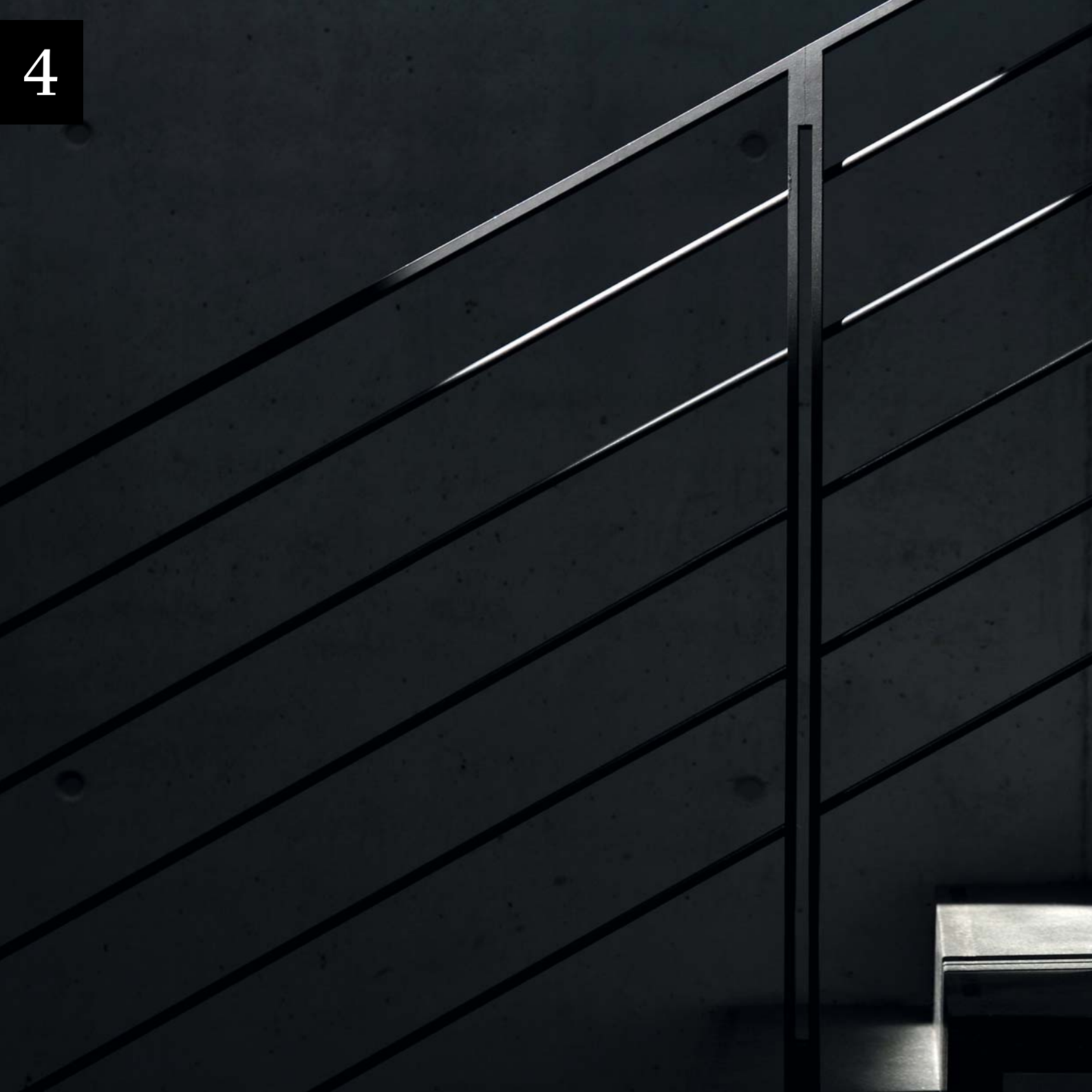
A gradient echo is generated by switching gradient pulses of reversed polarity.

The echo time has to be short because the gradient echo can be generated only within the T_2^* decay.

The gradient-echo technique is typically faster than the spin-echo technique.


4

A Study of Contrasts

MR imaging is unique in the way it can control image contrast and therefore expand the diagnostic range. The art of MR application lies in the choice of pulse sequences and the combination of acquisition parameters.

Spin echoes and contrast weighting

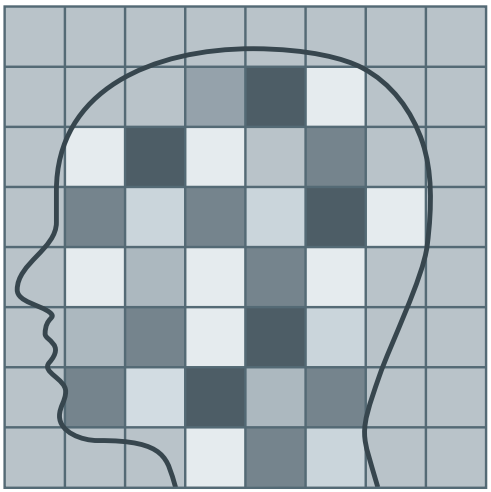
Demonstrating the three most important types of contrast in MR imaging: T_1 contrast, T_2 contrast, and proton density contrast

What determines image contrast?

How do we obtain an image with the largest possible contrast between different tissue types? Different tissue types have different transverse magnetizations. Where the signal is strong, the image shows bright pixels; weaker signals result in darker pixels.

What determines the signal strength? Clearly to a large degree, the proton density in the respective voxel: the greater the number of protons contributing to the magnetization, the stronger the signal.

But even more important for medical diagnostics is the effect of the two relaxation constants T_1 and T_2 on the image contrast.

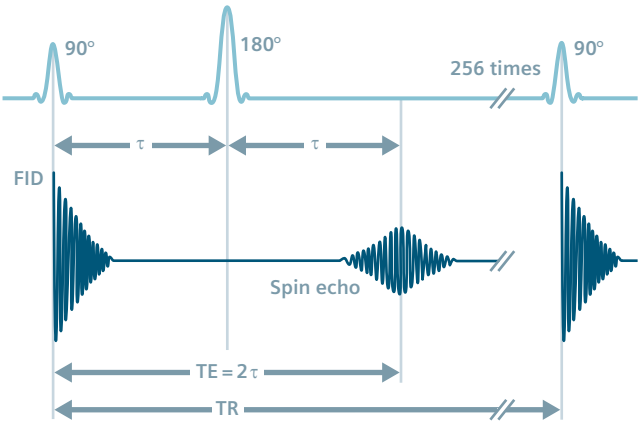


Important parameters: TE and TR

Remember the spin-echo sequence? A 180-degree pulse is applied at time τ after a 90-degree RF pulse, a spin echo is generated after echo time $TE = 2\tau$.

This pulse sequence, 90 degrees—180 degrees has to be repeated until all phase-encoding steps of the scan matrix have been acquired (for example, 256 times). The time interval between the repetitions is called **repetition time** TR.

TE and TR are important parameters for controlling the contrast of a spin-echo sequence. Let us follow these two temporal parameters and see how they affect image contrast.



Proton-density contrast

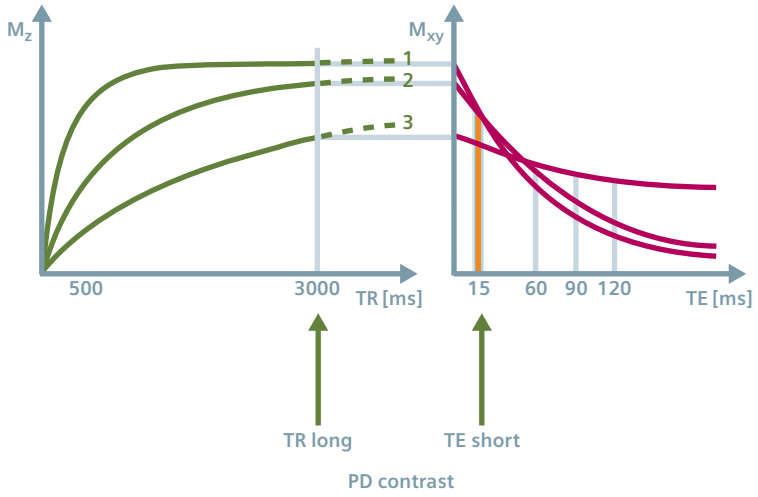
Based on differences in the number of protons per voxel

The maximum value of magnetization

The diagram shows three different tissue types (1, 2, 3) with different relaxation times.

Longitudinal relaxation begins immediately after the 90-degree pulse. The longitudinal magnetization M_z of the three tissue types recovers at different speeds. Their maximum values correspond to the **proton densities**, that is, the number of hydrogen protons per volume unit.

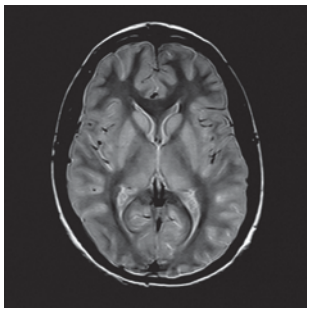
By means of a repeated 90-degree pulse after repetition time TR, the actual longitudinal magnetizations are converted into transverse magnetizations M_{xy} and generate signals of different strengths.



If we select a sufficiently *long* repetition time TR, the difference in signal in the tissue after a repeated 90 degree pulse depends mainly on the proton density of the tissue because of the nearly complete longitudinal relaxation.

Should we decide to generate echoes very shortly after the repeated 90-degree pulses, that is, with a *shorter* echo time TE, we will obtain a proton density-weighted image (PD for short).

In actual application, the TR of a spin-echo sequence is rarely longer than two to three seconds. However, this also means that tissue types with longer T_1 constants, for example, CSF, will not have recovered completely after this time period.



Proton-density contrast: TR long (2,500 ms), TE short (15 ms). The greater the proton density of a tissue type, the brighter it appears in the PD image.

T_2 contrast

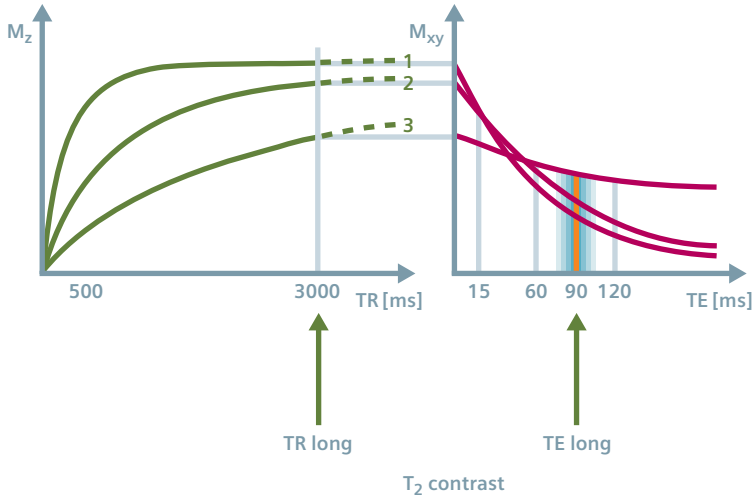
Based on the speed of transverse relaxation

Signal strength depends on T_2 decay

Let us stay with the *long* repetition time TR. What happens when we also select a *long* echo time TE?

The signal curves decrease due to T_2 relaxation and even might begin to intersect. The proton density contrast is lost. With longer echo times, the curves begin to diverge and the contrast is controlled by the T_2 relaxation. We obtain a T_2-weighted image.

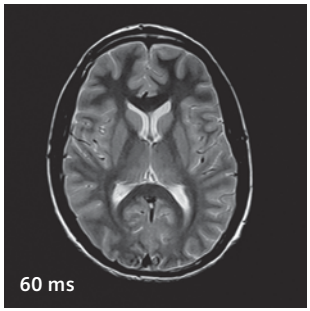
The signal strength of the spin echo typically depends on the T_2 decay.

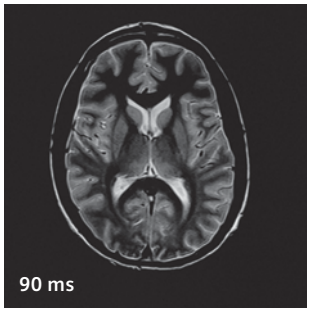


The comparison of images shows T_2 contrast with increasing echo time TE.

As the echo time increases, at some point the proton density no longer influences contrast. The T_2 contrast depends strongly on the TE selected. The optimal TE of a T_2-weighted image is the mean value of the T_2 constants of the tissue to be displayed (in our case between 80 ms and 100 ms).

If the echo time is too long (last image), transverse magnetization decays to a level where the signal of some tissue types disappears in the unavoidable signal noise.





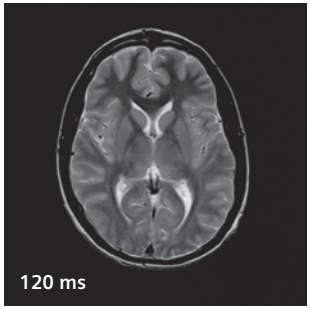


Image comparison with respect to T_2 contrast: TR long (2,500 ms), TE is increasing. CSF with a long T_2 appears bright in a T_2-weighted image.

T_1 contrast

Based on the speed of longitudinal relaxation

Signal strength depends on T_1 build-up

What happens when we select a *short* repetition time TR so that the T_1 relaxation is not complete. The signals will be much weaker and the contrast will decrease rapidly with increasing echo time. For this reason, we have to select the *shortest* possible echo time TE.

A short TR reduces the effect of the proton densities, a short TE cancels the effect of T_2 relaxation. The difference in signal strengths depends largely on the previous longitudinal magnetizations, that is, from the T_1 relaxation of the tissue. We obtain a T_1-weighted image.

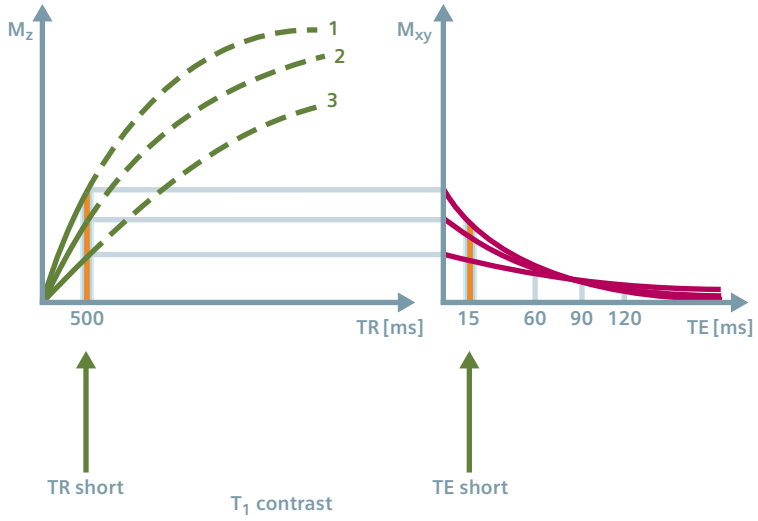
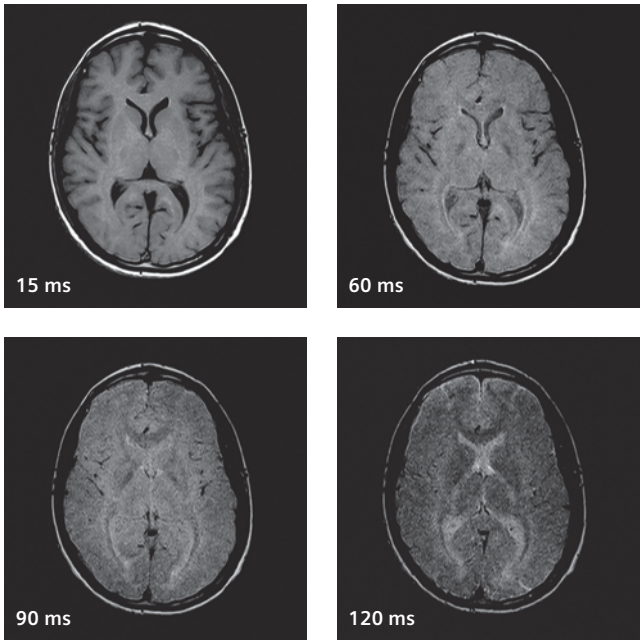


Image comparison with respect to T_1 contrast: TR short (500 ms); TE is increasing. CSF with a long T_1 appears dark in a T_1-weighted image. The optimal TR corresponds approximately to the average T_1 constant of the tissue type to be displayed. This means between 400 ms and 600 ms for 1.0 to 1.5 tesla.

The image comparison shows good T_1 contrast when TR and TE are both short.

With longer echo times, both the T_1 contrast and the measurable signal are reduced. The combination of short repetition time and long echo time is obviously completely unsuitable.

Normal soft tissue types differ only slightly in their proton density. However, they do show different T_1 relaxations. For this reason, T_1-weighted imaging is highly suitable for anatomical displays.

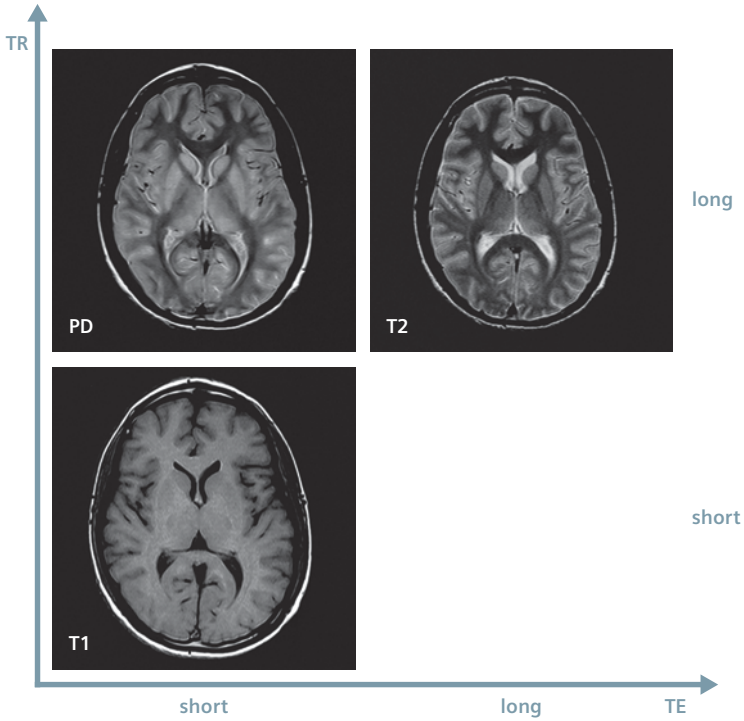


REDUCED TO THE ESSENTIALS

The comparison of images shows the three important combinations of TR and TE as well as their resulting contrast weighting:

- T_1 contrast (TR short, TE short)
- T_2 contrast (TR long, TE long)
- Proton density contrast (TR long, TE short)

In spin-echo imaging, the effects of T_1 and T_2 are reversed: tissue with longer T_1 appears *darker* in the T_1-weighted image, tissue with longer T_2 appears *brighter* in T_2-weighted images.



Contrast with inversion recovery

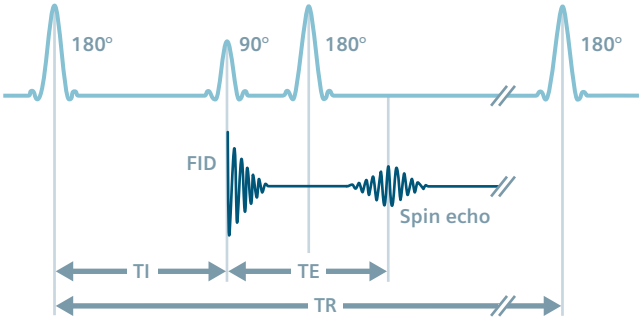
A spin-echo sequence with a 180-degree preparation pulse

Inversion time determines contrast

The inversion recovery sequence (IR) is a method for creating a signal depending primarily on T_1. The IR sequence applies pulses of 180 degree—90 degree—180 degree. The longitudinal magnetization is first flipped by the 180 degree **preparation pulse** in the opposite direction—that is, it is inverted. Transverse magnetization is therefore zero and we do not receive an MR signal.

The interval between the 180-degree pulse and the 90-degree stimulation pulse is known as **inversion time** TI. During this time period, the longitudinal magnetization recovers.

The 90-degree excitation pulse converts the momentaneous longitudinal magnetization into transverse magnetization.

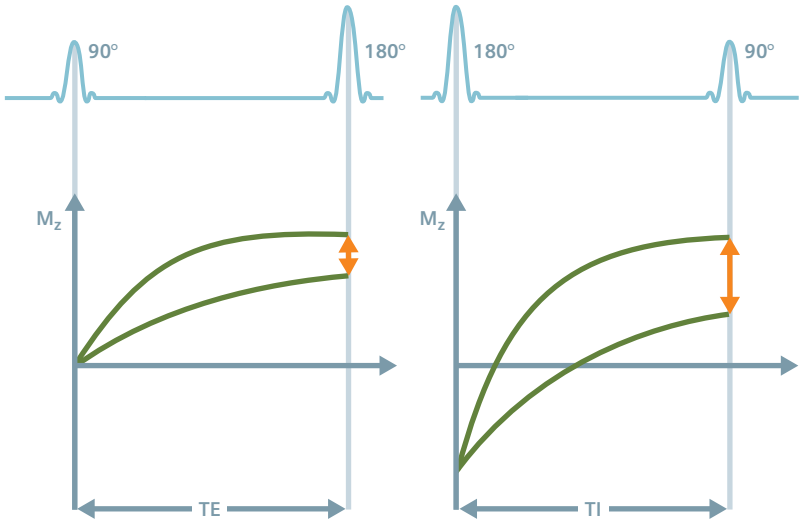


Strong T_1 contrast

While the standard spin-echo sequence provides for T_1 contrast and in particular an excellent T_2 contrast, the inversion recovery sequence is known for its higher T_1 contrast.

As the longitudinal magnetization relaxes its negative value following inversion, the magnetization of different tissue types reaches zero at different times. The inversion of the magnetization improves the dispersion of these TI curves, leading to better T_1 contrast. By selecting a suitable inversion time TI, the contrast is optimized.

We can use IR sequences to display the most minute T_1 contrasts, for example, in the brain of a newborn. The disadvantage is the longer measurement time. Depending on TI, fewer slices are measured than with a T_1-weighted spin-echo technique.



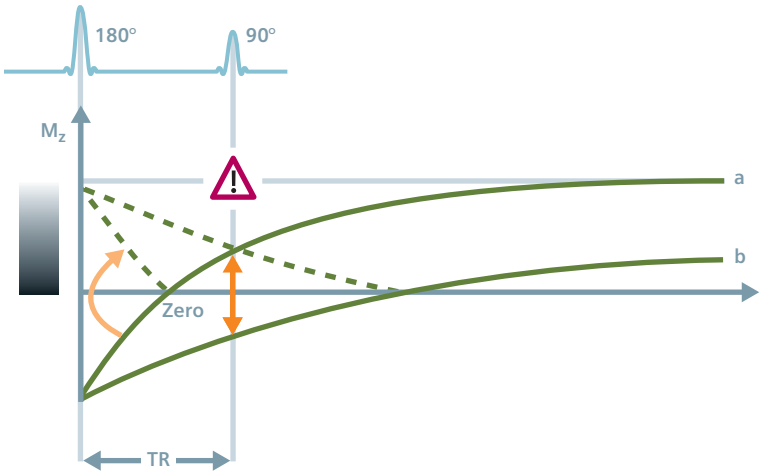
The zero point

Making tissue disappear

Gray on gray and the zero signal

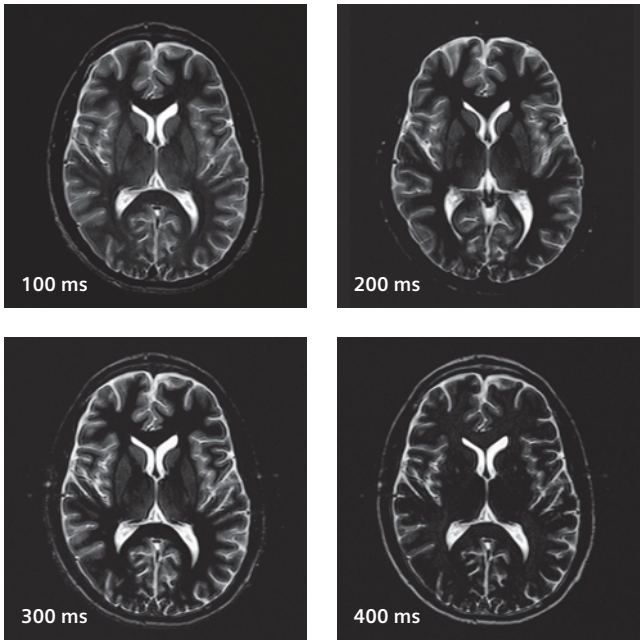
Let us look at the curves of the longitudinal relaxation for a special case. Because of the TI selected, the faster relaxing tissue (a) has already passed the zero point, while the slower relaxing tissue (b) has not.

This can be very confusing if only the magnitude of the signals is used for image contrast. No differentiation is made between the positive and negative longitudinal magnetization. Tissue types with widely different T_1 constants would be displayed with the same gray value.



The comparison of images shows the effect of inversion time TI on the contrast in the brain. The signals from white or gray matter may disappear.

Image comparison with respect to contrast with inversion recovery: TI is increasing. The signal of white matter decreases as the inversion time TI increases and goes through the zero point at TI = 300 ms. At TI = 400 ms, the signal of gray matter (with longer T_1) has reached zero point, while the signal of white matter is increasing again.



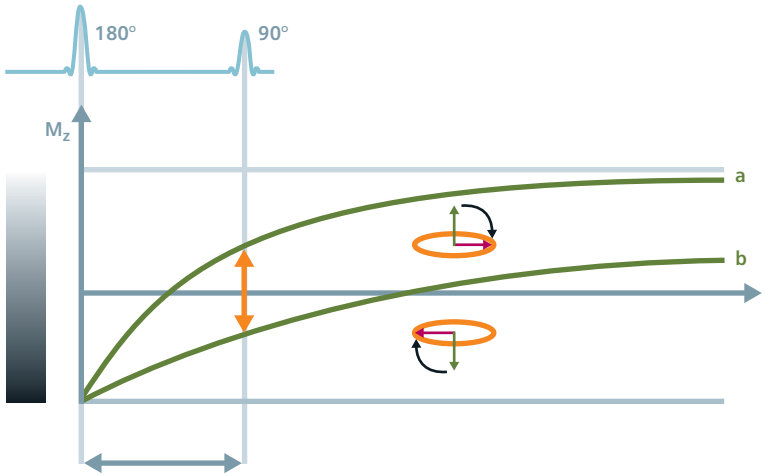
True inversion recovery

T_1 contrast all the way

Taking into account the sign of the magnetization

How can we ensure contrast between different tissue types? By considering the orientation of the longitudinal magnetization.

The positive and negative longitudinal magnetizations are converted by the 90-degree excitation pulse into transverse magnetization with a 180-degree phase shift. If we consider both the magnitude as well as the phase difference of the signals, it is possible to allocate the signals to the original positive or negative longitudinal magnetization. This will ensure maximum T_1 contrast.



This technique of phase-sensitive reconstruction gives the true longitudinal magnetization and is also known as **true inversion recovery**. Its preferred application is in the area of pediatrics.

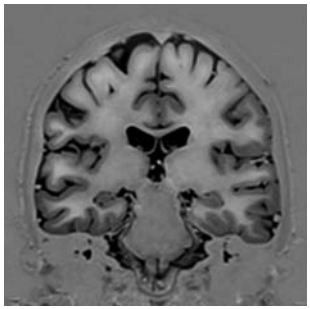


Image background, usually black, is shown in mid-range gray when phase-sensitive reconstruction is used.

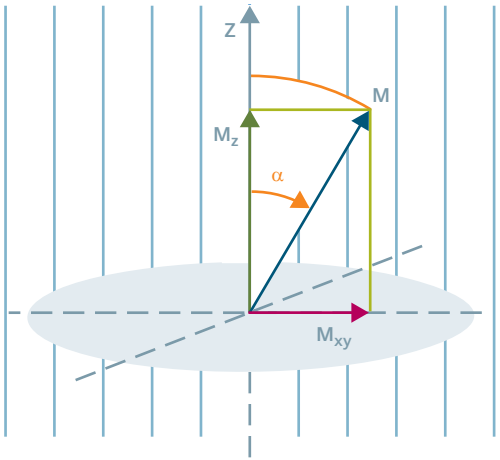
Contrast with gradient echoes

Increased signal and shorter measurement times with a reduced flip angle

Shortening the repetition time without signal loss

There are different types of gradient-echo sequences applied in clinical practice. A major benefit of a gradient-echo sequence is the possibility to decrease repetition time TR. The magic "trick" is the 'low-angle' tech-nique: a reduced flip angle α of less than 90 degrees. In this case, we do not have the effect of the entire available magneti-zation M in the xy-plane, but rather only part of it is converted into a transverse magnetization M_{xy}.

On the other hand, the longitudinal magnetization is *not* zero after such an α-**pulse**, but continues to have a reduced magnitude M_z.



Boost in signal

For example, an RF pulse with a flip angle of 20 degrees already generates a sufficiently high transverse magneti-zation of 34 % of its maximum value. In this case, the remaining longitudinal magnetization reaches 94 % of its maximum value.

At the next pulse, a high longitudinal magnetization is available again. At very short repetition times (less than T_1), At very short repetition times, after several 20-degree pulses, the magnetization will reach a 'steady state'. The generated MR signals are even larger than with 90-degree pulses at the same short TR. This boost in signal allows for very short repetition times and a greatly reduced measurement time.

The contrasts generated with a gradient-echo sequence rather than with a spin-echo technique are very complex.

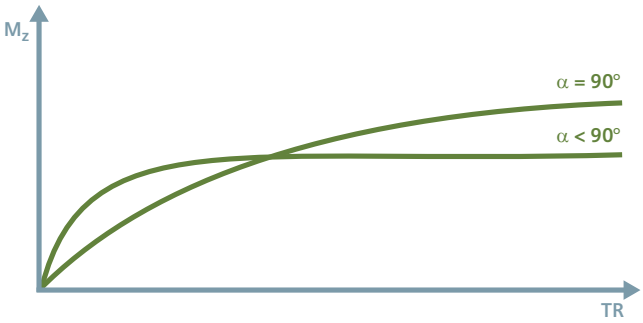
The optimal flip angle

Balancing the flip angle between T_1 and TR

Ernst angle and the steady state

For a tissue type with a specific T_1, a maximum signal is generated at a *defined* flip angle, the so-called **Ernst angle**. This optimal flip angle is a function of the selected repetition time TR.

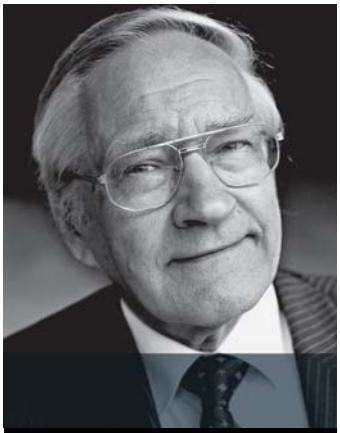
We know that the longitudinal magnetization recovers in proportion to its size (exponential growth). By flipping it by an angle α the remaining longitudinal magnetization is smaller than before (at 20 degrees, it is 94 % of 94 %, etc.). But it also recovers accordingly faster. After repeated α-pulses, an equilibrium is obtained between the opposing tendencies; the longitudinal magnetization remains the same after each pulse. This equilibrium is also known as **steady state**.



Calculating the Ernst angle

Let us calculate the optimal flip angle, for example, in the brain: With T_1 = 800 ms and a TR of 2500, an angle of 87 degree results which is not very different from 90 degrees. But with a TR of 100, the optimal flip angle is only 28 degress.

For proton-density weighted images, an angle smaller than the Ernst angle is desirable.



"The optimum angle for producing the maximum magnetization in the xy-plane is easily found. The average optimum flip angle is given by $\cos \alpha = \exp(-TR/T_1)$*"*

Richard R. Ernst

Swiss MR researcher Richard R. Ernst (Nobel prize 1992): "I am not aware of any other field of science outside of magnetic resonance that offers so much freedom and opportunities for a creative mind to invent and explore new experimental schemes that can be fruitfully applied in a variety of disciplines."

FLASH

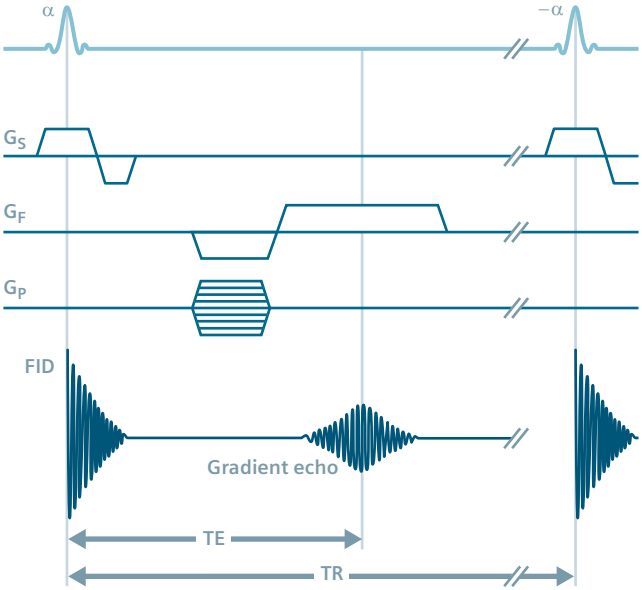
Spoiling the transverse magnetization on purpose

Exploiting the steady state of longitudinal magnetization

The FLASH gradient-echo sequence (**F**ast **L**ow **A**ngle **Sh**ot) uses the equilibrium of longitudinal magnetization.

With a very short repetition time TR, we have a residual transverse magnetization prior to each emitted α-pulse. This remaining magnetization would contribute to the next acquired signal, in a FLASH sequence it is typically eliminated by a strong "spoiler" gradient pulse.

RF spoling can also be done by applying randomized phases of the subsequent α-pulses.



Contrasts with FLASH

FLASH allows for very strong T_1 contrasts using an extremely short TR.

- T_1 contrast:
  TR short (40–150 ms)
  TE short (5–10 ms)
  α medium to large (40°–80°)
- T_2^* contrast:
  TR long (500 ms)
  TE relatively long (18–40 ms)
  α small (5°–20°)
- Proton density contrast:
  TR long (500 ms)
  TE short
  α small (5°–20°)

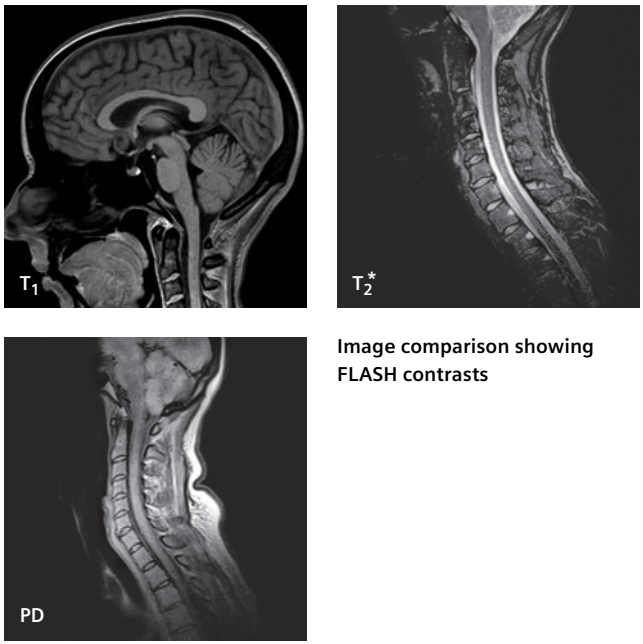


Image comparison showing FLASH contrasts

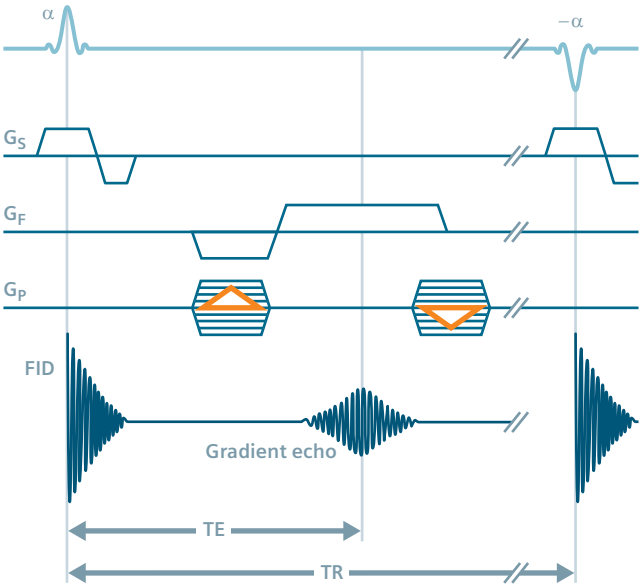
FISP

Utilizing a uniform transverse magnetization

Refocused gradient echoes

With the FISP sequence (Fast Imaging with Steady-state Precession), the remaining transverse magnetization is not eliminated before the next α-pulse. Instead, it contributes to the signal along with the longitudinal magnetization.

To obtain a uniform transverse magnetization, the dephasing gradients in the phase-encoding direction (G_P) are compensated after the echo by applying gradients of reversed polarity. The magnetization is flipped in the opposite direction after each repetition time TR with a negative α-pulse. Subsequent α-pulses have alternating polarity.

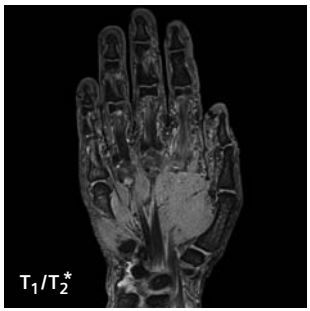


Contrast with FISP

The strength of the longitudinal magnetization depends on T_1, the amplitude of transverse magnetization depends on T_2^*. Contrast is a function of the ratio of T_1 to T_2^* and is generally not dependent on TR.

- T_1/T_2^* contrast:
  TR short
  TE short
  α medium

Repetition time TR should be as short as possible. A long TR makes FISP behave like FLASH.



Contrast with 3D FISP


5

The Strategy of Image Quality

Image quality is most important in MR imaging. The secret lies in optimizing image quality in relation to the measurement time required.

Contrast, signal, and noise

The most important criteria for image quality: a strong signal, low noise, good contrast, as well as sufficient resolution.

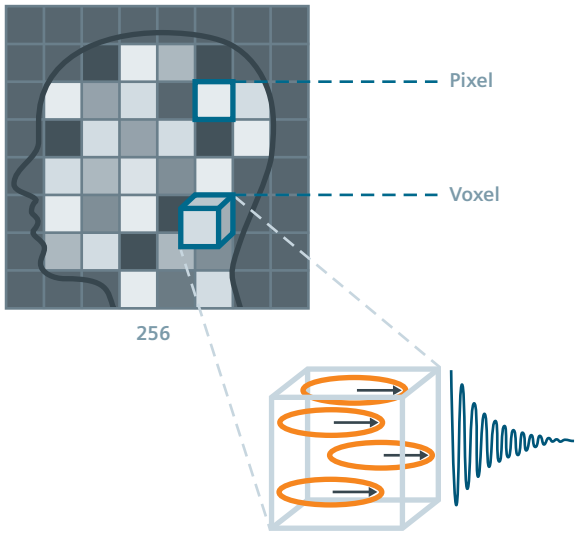
Let us start with a strong signal

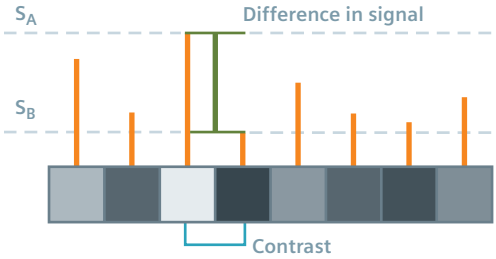
The MR image consists of a multitude of image pixels. Each pixel has a specific gray value. The pixels in the image represent the individual volume elements (voxels) in the slice measured.

During measurement, each voxel in a slice contributes to the overall MR signal.

First and foremost, the **signal strength** per voxel depends on the quantity of signal-generating protons, i.e. the local proton densities.

The more protons contribute to magnetization, the stronger the signal.





This is how contrast is generated

Hyperintense pixels in the image represent stronger signals, weaker signals result in hypointense pixels. For simplicity's sake, our example shows a single row of eight pixels.

The **contrast** in the image is the relative difference in the signal strength between two adjacent tissue types, A and B. In other words, contrast is equal to the difference in signal:

Contrast = Difference in signal = $S_A - S_B$

Contrast resolution is the ability to distinguish differences in grayscale colors of adjacent structures. Each type of tissue emits individual signal strengths. This allows for anatomical differentiation in the image and, in the final analysis, differentiation between pathalogical and healthy tissue.

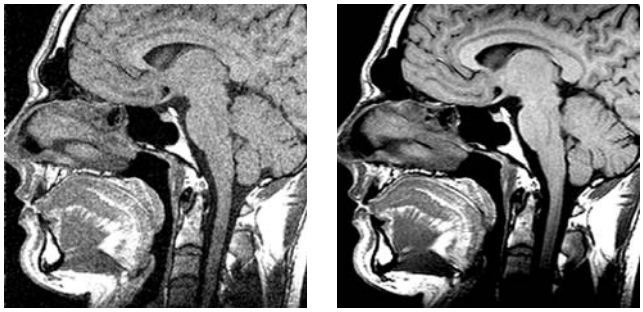


Image comparison: Low T_1-contrast (left), high T_1-contrast (right)

Signal versus noise

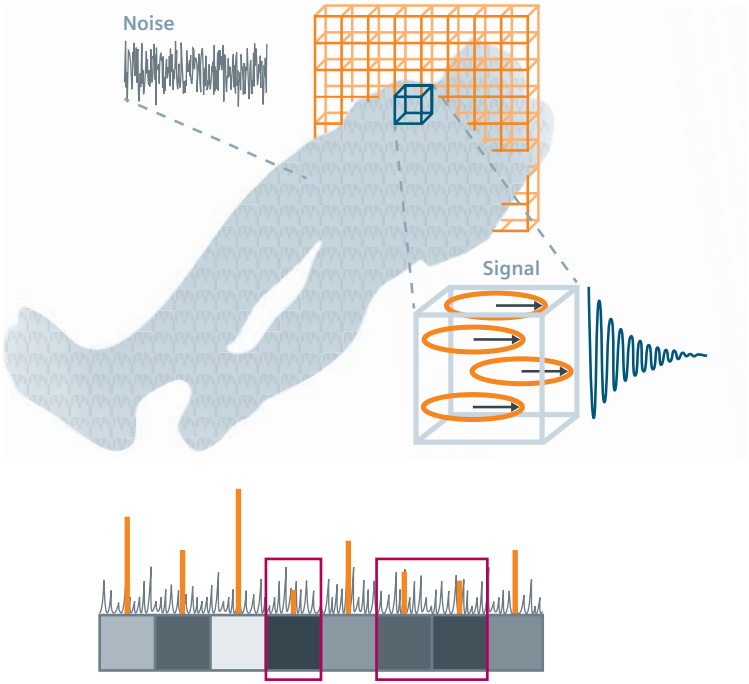
Noise is unavoidable but has a strong adverse affect on image quality

If noise takes over

The NOISE in the image appears as a grainy, random pattern similar to snow on a TV screen. It represents statistical fluctuations in signal intensity that do not contribute to image information. There are basically two sources for this effect.

Noise is generated throughout the human body through Brownian motion of molecules. To that we add the electronic noise of the receiver technology.

We are faced with a problem when the signal from a slice is too weak. In this case, the signal may be "washed over" permanently by noise.

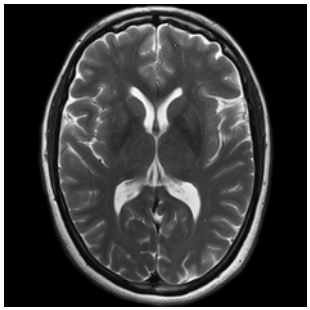


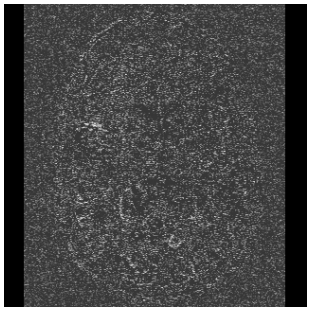
The signal-to-noise ratio

An important criterion for MR image quality is the relationship between the intensity of the wanted signal and the statistical noise, the **signal-to-noise ratio** (SNR):

$$SNR = \frac{Signal}{Noise}$$

A higher SNR means a less grainy image.





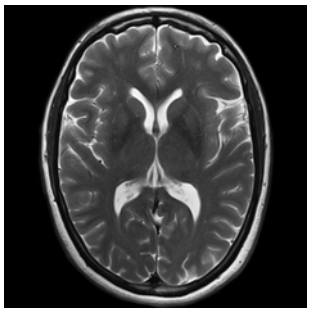


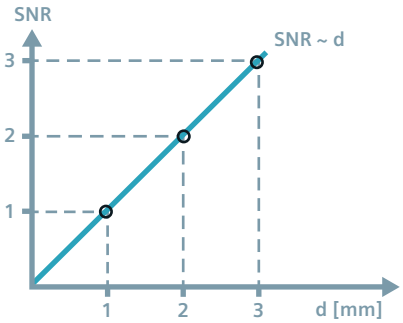
Image comparison:
The two images on the left were acquired in the same way and subsequently subtracted from one another (= pixel-by-pixel difference in gray values). What remains is background noise (on the right).

Amplifying the signal

A strong signal as the first step toward good image quality

Increasing slice thickness

Let us assume we enlarge the voxel by measuring a thicker slice. As a result, **signal intensity** increases, since more protons are contributing to signal strength. The show-stopper is: The portion of noise remains the same since it is not just coming from the slice, but rather overall from the patient's entire body (or more precisely, from the sensitive volume of the receive coil). Also: the thicker the slice, the stronger the signal. And the higher the SNR.



The SNR is directly proportional to the voxel size.

Disadvantages: Increasing slice thickness reduces spatial resolution in slice direction. This may result in partial volume effects that distort image results (for example, bones protruding into soft tissue).

Please note: Doubling the matrix size to im-prove the in-plane resolution will decrease the voxel size by a factor of four. Consequently, SNR will be decreased by a factor of four as well.

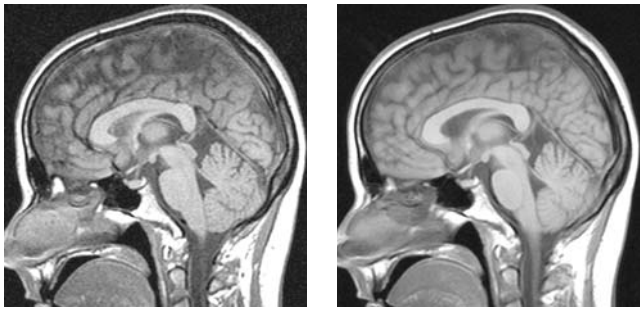
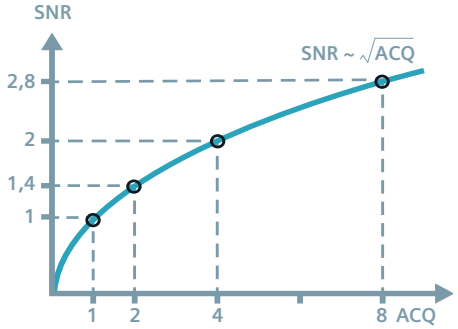


Image comparison: The figure on the right shows a slice that is three times as thick as the slice in the left image. As a result, the SNR is three times as high.

Increasing the number of acquisitions

We do not have to select an undue large slice thickness. The SNR can be improved by other methods as well: by measuring one slice several times (several **acquisitions**) and by adding the results in a single image.

However, the SNR is not increasing linearly, it is rather getting to be less.



SNR is proportional to the root of the number of acquisitions.

For example: If four acquisitions are measured and averaged on one slice, the total SNR is twice what is was before.

Disadvantage: The measurement time increases as the number of acquisitions increases.

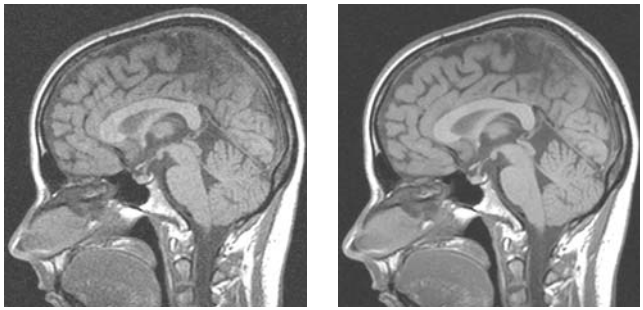


Image comparison: left 1 measurement, right 4 measurements. Results: The SNR on the right is double the SNR on the left.

The effective contrast

A quality criterion that relates to what we see in the image

Combining contrast and noise

The contrast we actually see and evaluate in the image is more than differences in signal intensity: This effective contrast is also related to the noise level.

A high signal-to-noise ratio (SNR) alone does not guarantee easy differentiation between two structures in an image.

The **contrast-to-noise ratio** (CNR) in an MR image is the difference between the signal-to-noise ratios between two tissue types, A and B:

$CNR = SNR_A - SNR_B$

Since CNR equals effective contrast, it is a better quality criterion than SNR.

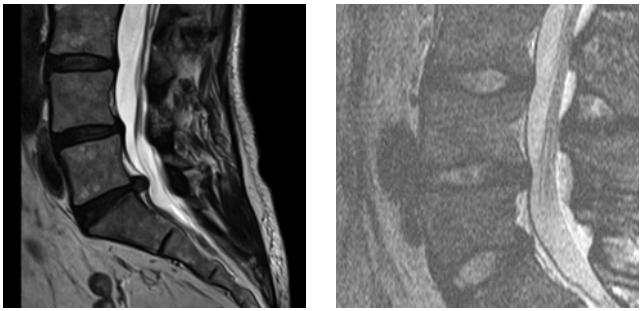


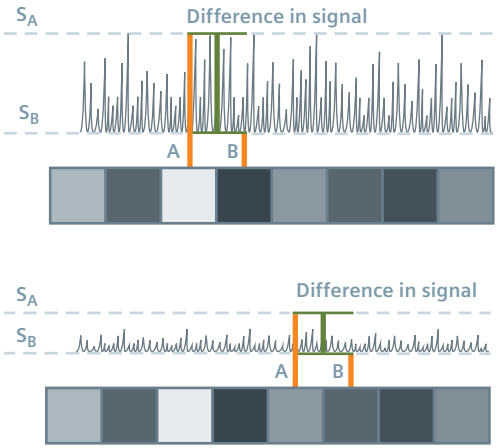
Image comparison: good CNR, poor CNR

When contrast is too noisy

Let us assume we have a noticeable difference in signal between two tissue types A and B. In this case, we could obtain good contrast. However, if we set this difference in signal in relation to high noise, the contrast drowns in noise.

Our example: Although the difference in signal is higher in the first than in the second case, the CNR and consequently the effective contrast is lower.

To obtain good image quality, the difference in signal between two types of tissue has to be significant despite the noise.



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SNR and CNR are important criteria for MR image quality. They relate signal and contrast to noise.

The signal strength is in part determined by the proton density in the respective voxel. The more protons contribute to magnetization, the stronger the signal.

Contrast is the difference in signal strengths between two relevant tissue types.

The signal-to-noise ratio (SNR) describes the relationship between the intensity of signal and noise . SNR can be improved by increasing both the slice thickness as well as the number of acquisitions.

The difference of the signal-to-noise ratios between two types of tissue is the contrast-to-noise ratio (CNR). It is a better quality criterion compared to SNR.

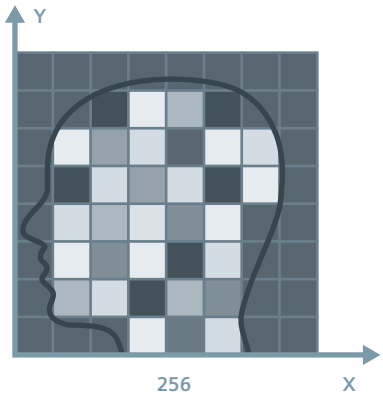
About image size and resolution

Controlling the spatial resolution of the structures acquired in the image

The matrix determines resolution

The matrix size determines resolution and measurement time. If you halve the number of rows (the phase resolution), you also halve the measurement time, since the number of time-consuming phase-encoding steps is also reduced by 50%. If you use twice the phase resolution, the measurement time accordingly increases by 50%.

Measurement time = phase resolution ×
× TR (repetition time) × number of acquisitions



Example: For a phase resolution of 256 sampling points, 500 ms TR and one acquisition, the measurement will be completed in 128 seconds.

Matrix size and signal-to-noise ratio

The matrix size also affects the signal-to-noise ratio. Because with larger voxels, more protons contribute to the signal.

If the measurement matrix is enlarged without changing the other parameters, resolution will increase accordingly. The voxels become smaller and thus SNR is reduced as well. SNR is proportional to the size of the voxel. This means that at a constant slice thickness, SNR is proportional to the pixel size.

Matrix	Relative SNR
128	4.0
256	1.0
512	0.25

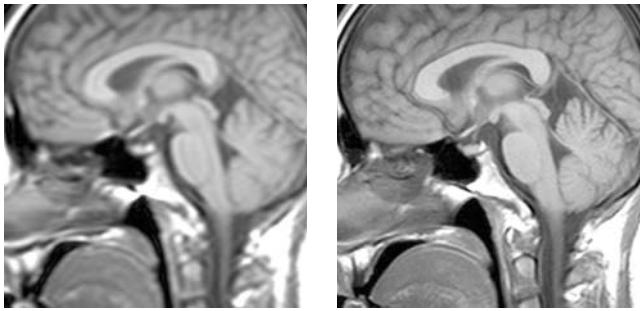


Image comparison: Matrix 128 (left) low resolution and better SNR, Matrix 256 (right) higher resolution and decreased SNR

The field of view

Optimizing in-plane resolution and measurement time

What is the field of view?

The **field of view**, abbreviated to FOV, is the square or rectangular image area to be measured (in mm). In short, the FOV determines what you can see in the MR image.

To save time and to obtain maximum resolution, the FOV is optimally adjusted to the area under examination.

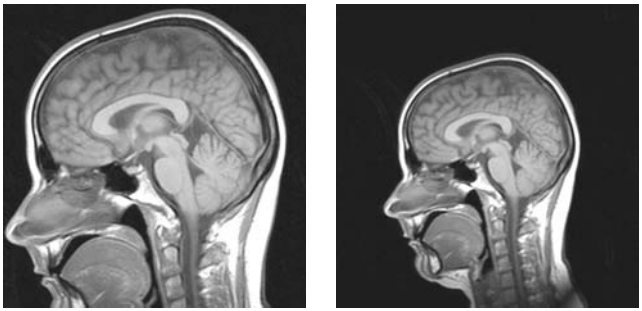


Image comparison: FOV = 230 mm (left), FOV = 330 mm (right) is unnecessarily large

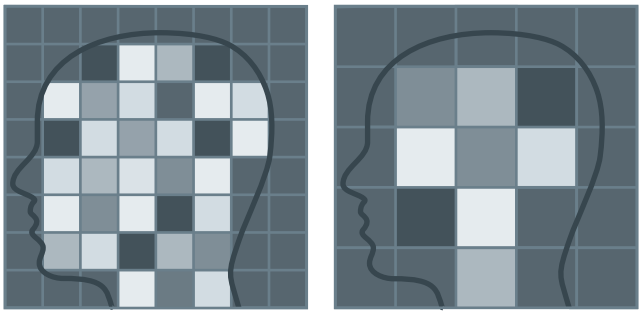
Pixel size and in-plane resolution

The smaller the field of view at a fixed matrix size, the higher the in-plane resolution. The number of pixels per in-plane unit increases while the pixels themselves decrease in size. Conversely, for a given matrix size and larger FOV, the pixels are greatly enlarged. The resolution is reduced accordingly.

$$Pixel\ size = \frac{FOV}{Matrix\ size}$$

Smaller pixels mean an improved in-plane resolution.

FOV (mm)	Matrix size
256	256×256
256	128×128
128	128×128



FOV 256 mm
Matrix 256x256

FOV 350 mm
Matrix 256x256

Right image is magnified to show the same size of the head. Pixels are larger, resolution is lower.

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Matrix size, field of view (FOV) and slice thickness affect resolution as well as measurement time and the signal-to-noise ratio.

Changing these parameters has a number of effects. The most optimal solution is a compromise—primarily between image quality and measurement time.

Summary of the effects of the parameters:

	Measurement time	Resolution	SNR
Matrix ↑	↑	↑	↓
FOV ↑	–	↓	↑
Slice thickness ↑	–	↓	↑

Speeding up the measurement

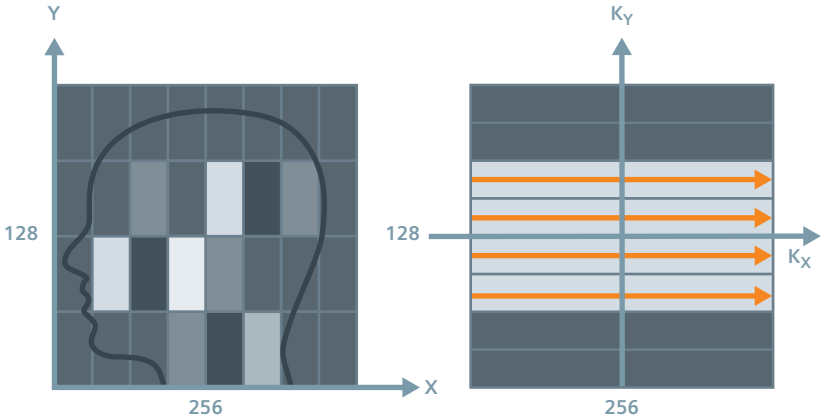
Reducing measurement matrix and field of view for a faster measurement

Decreasing phase resolution

To accelerate the measurement, you can select a reduced measurement matrix with a lower phase resolution, for example, instead of 256×256 you select 128×256. The pixels are now rectangular.

Why is the measurement faster? The phase resolution of the measurement matrix corresponds to the number of phase-encoding steps (NP). This means it is directly proportional to the measurement time (NP×TR).

A halved phase resolution (for example, 128) corresponds to half the number of phase-encoding steps. The measurement time is halved.



Filling the raw-data matrix

As the reduced measurement matrix is rectangular, signals are acquired only for the central raw data lines. The missing outer rows of the square k-space are filled with zeroes.

Why does this work? As you might remember, fine structures are shown in the outer regions of k-space. The central rows provide the important contrast. When an image is reconstructed from the raw data, the image pixels are interpolated in the phase-encoding direction.

Image resolution is reduced with phase resolution (for example, phase resolution reduced by 50%, image resolution by 50% in this direction). Since the voxels are larger, SNR is improved.

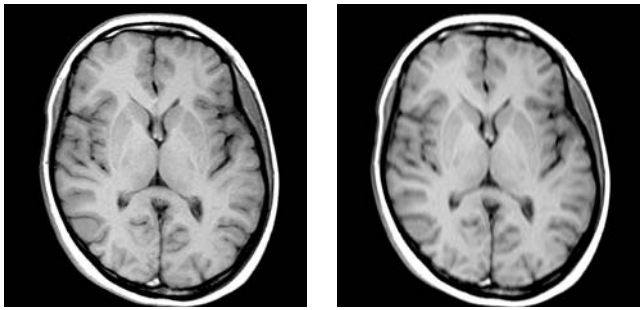


Image comparison:
Phase resolution 100% (left) and 50% (right)

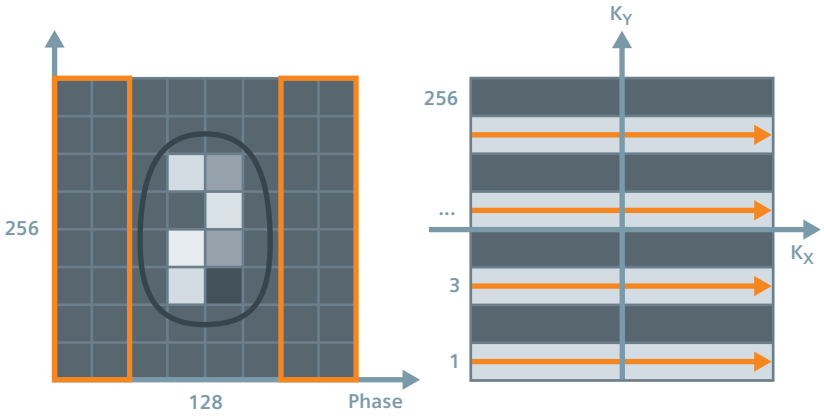
Cutting the field of view

Cutting the FOV in phase-encoding direction accelerates image acquisition

Rectangular field of view

If the object to be measured does not fill a square image, we can select a rectangular field of view. If we cut halve the FOV in the phase-encoding direc-tion, we only need half as many phase-encoding steps. Only every other row is filled with raw data, the others contain zeroes only.

The measurement time is directly pro-portional to the number of phase-encoding steps. For this reason, the measurement time is reduced by 50% with half the FOV.



Measuring faster at the same resolution

At half the FOV and half the number of phase-encoding steps, the voxel size remains unchanged and so does the resolution.

SNR is decreased.

So a rectangular FOV is an acceptable choice for accelerating image acquisition.

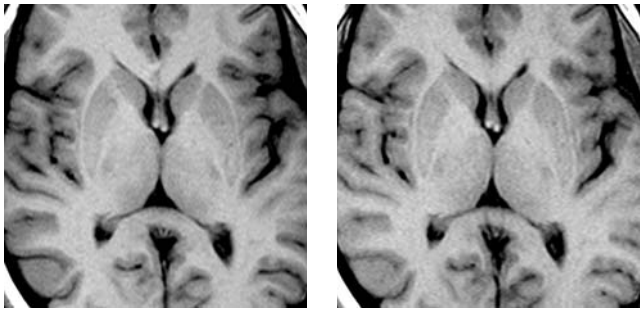


Image comparison:
FOV Phase 100% (left) and 50% (right)

Measuring only part of the raw data

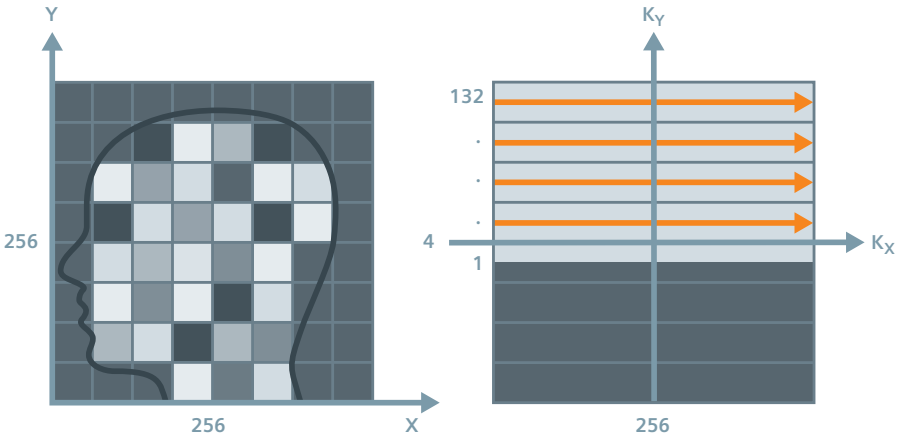
Utilizing the symmetry of k-space

Half-Fourier

With the half-Fourier technique, only half of the raw data matrix (k-space) is filled with data in the phase-encoding direction. The missing data are reconstructed symmetrically.

Unavoidable small magnetic field inhomogeneities lead to phase distortions. For this reason, slightly more than half of the phase-encoding steps are acquired for phase correction.

As a result, the measurement time is almost cut to half.



Partial raw data

The partial Fourier technique works in the same way as the half-Fourier technique: Only part of the k-space is filled in the phase-encoding direction (5/8, 6/8, or 7/8).

What is the image quality like? Because the voxel size has not changed, the resolution has not deteriorated. SNR decreases. In most cases, there will be a barely discernable difference between images with and without half- or partial Fourier.

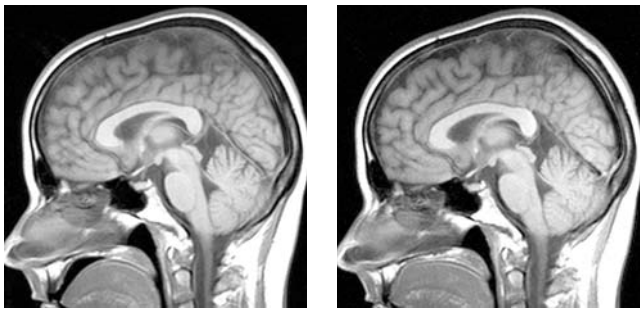


Image comparison: Normal (left) and half-Fourier (right)

Summary

The measurement time is shortened by reducing the phase resolution or both the phase resolution and the field of view. The examples provided demonstrate the effects on SNR and resolution:

	SNR	Resolution
Field of view 100% Phase resolution 50%	Better	Less
Field of View 50% Phase resolution 50%	Less	Unchanged
Half Fourier	Less	Unchanged

Increasing Diagnostic Value

Since the invention of MRI, progress has never stopped. Let us introduce some small, some big, some “old” and some new techniques and innovations that help to enhance the diagnostic value of MRI.

Packing echoes and slices

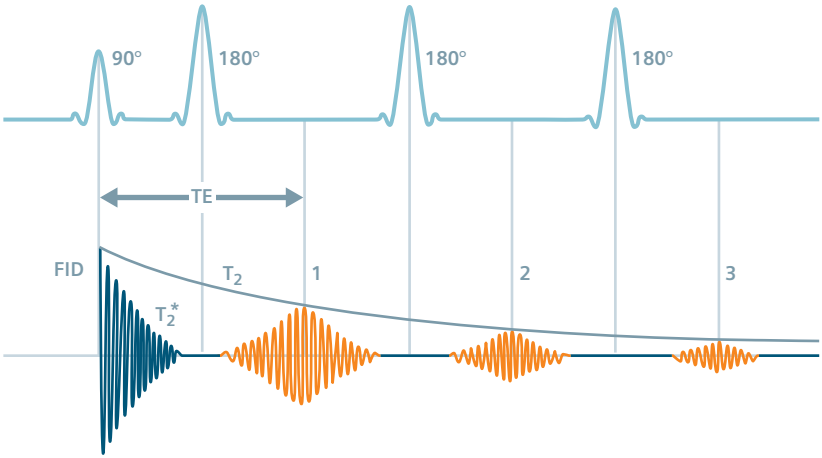
A multiecho experience

Multiple echoes with different degrees of T_2 weighting

Echoes following one another

We have discussed the generic pulse sequence for image acquisition in MRI. There is no reason to generate only one echo within the repetition time TR. In a **multiecho sequence**, several 180-degree pulses follow each other in sequence, thus several spin echoes with different degrees of T_2 weighting are generated.

Signal height of a multiecho sequence reduces with transverse relaxation: The longer the echo time, the smaller the echo. The echo time-dependent signal decay follows the T_2 relaxation time of the tissue. We can repeat this until transverse magnetization is irrevocably decayed.



Creating a pure T_2 image

The echo time-dependent drop in signal of a multiecho sequence can be used to calculate a pure T_2 image, a so-called **T_2 map**.

Multislice imaging

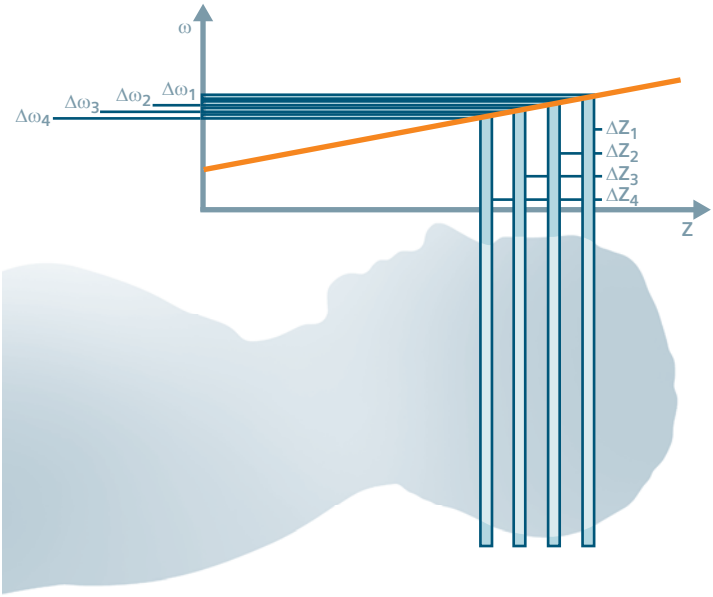
The standard in MRI

Measuring several slices at once

In order to cover a body region, we can measure a series of slices in 2D, or alternatively a 3D cube (see next section).

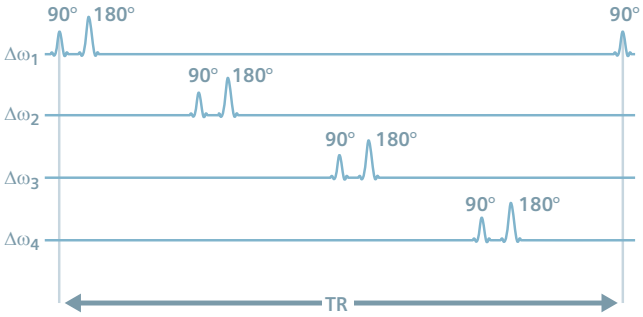
The linear slice-selection gradient of a pulse sequence allows us to distinguish different slices which can be excited by RF pulses in quick succession.

As echo time TE is always considerably shorter than repetition time TR, we can excite additional slices (for example, z_1 through z_4). by using a **multislice sequence**.



The multiecho sequence uses a series of 90-degree and 180-degree pulses with varying resonance frequencies that fit with the defined slice positions. This method provides us with all the slices necessary for examining a particular area or region during a measurement.

These slices do not need to be in straight axial orientation. By switching the gradient fields properly, we can scan slices with arbitrary orientations, even varying across slices.



Volume imaging: 3D

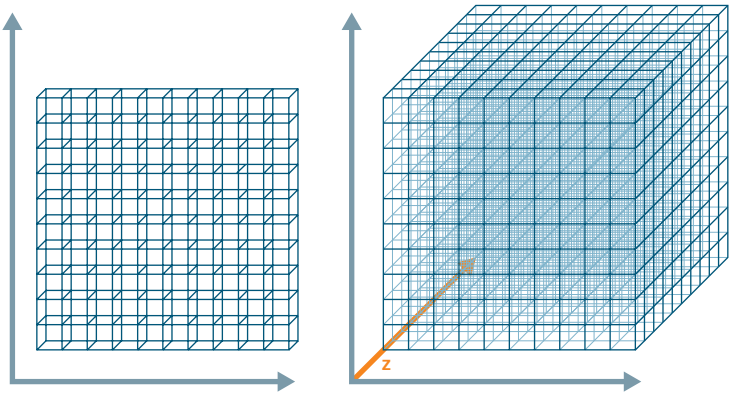
Creating spatial views from a region of interest

Generating 3D data

Fast 3D pulse sequences with short repetition time TR allow acquisition of three-dimensional data. The entire measurement volume, the **3D slab**, is excited and not just single slices.

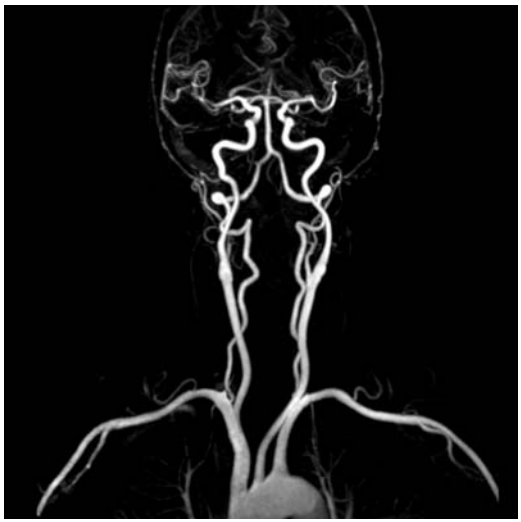
Different phase positions can be accurately located in space. This is the basic principle of phase encoding. In order to generate 3D data, we superimpose an additional phase-encoding gradient in the direction of slice selection ("z" in our example).

Through additional phase encoding perpendicular to the image plane and contiguous images, we obtain information regarding the defined slab. The planes of this volume are known as **partitions**.



Creating spatial views

With the help of standard post-processing software available, the 3D data sets generated can be used to create spatial views (for example, from the vessels).



Spatial image reconstructed from 3D data, a Maximum Intensity Projection (MIP)

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A mulitecho sequence generates multiple echoes by applying several excitation pulses in sequence, This is possible as long as T_2 relaxation continues.

Multislice imaging (2D): A multislice sequence creates a series of slices for a region of interest.

Volume imaging (3D): Fast 3D pulse sequences allow acquisition of 3D data sets from which spatial views can be created.

Take the echo train: Turbo spin echo

Shorter measurement times and improved tissue contrast

The emergence of turbo spin echo

Conventional spin-echo techniques can take several minutes per slice. Turbo spin-echo sequences (TurboSE, TSE) considerably shorten acquisition time and have replaced spin-echo techniques to a large extent. The use of TurboSE imaging has become routine in MRI today.

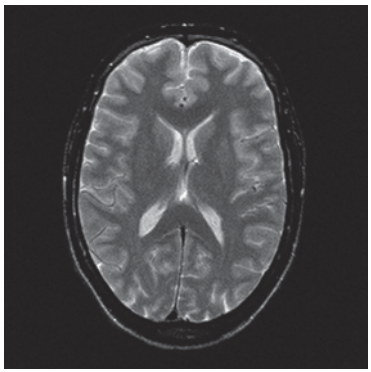
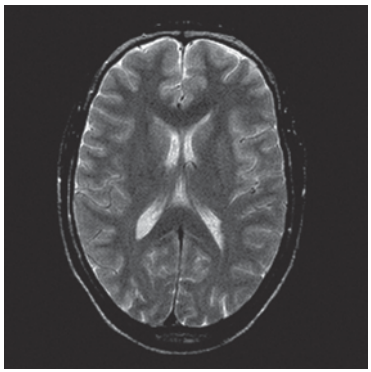


Image comparison: T_2 spin echo



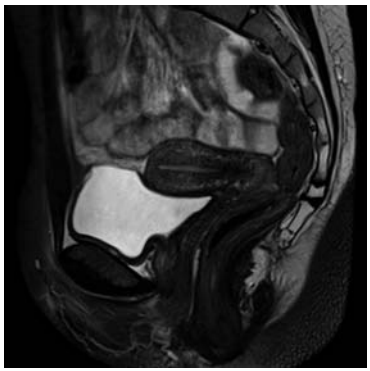
T_2 turbo spin echo

Turbo spin echo is based on RARE imaging, invented by the German clinical MRI researcher Jürgen Hennig in 1986: "The world (and the world of MR in particular) is full of 'impossible' things which turned out to become reality."

Strong in T_2 contrast

In the majority of cases, TurboSE sequences are used for T_2-weighted imaging. The most noticeable difference between TurboSE and spin-echo techniques is the bright fat signal in strongly T_2-weighted images. A T_1-weighted TurboSE technique is possible also, frequently used for imaging the spine.

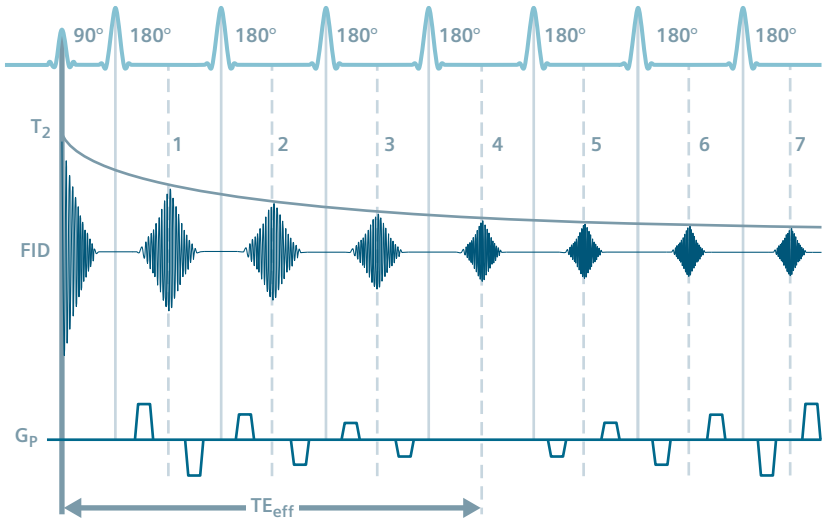
TurboSE sequences offer a far better contrast between white and gray matter. It is difficult to imagine neuroradiological imaging without the high-resolution possibilities offered by TurboSE sequences.



The turbo factor defines speed

TurboSE is a fast multiecho sequence: During the time it takes for a spin-echo sequence to acquire just one single echo, the TurboSE sequence acquires an entire series of echoes, known as the **echo tr**ain. Each echo of the echo train is given a *different* phase encoding (G_P) and fills one row of the raw-data matrix. The length of the echo train determines the maximum time savings or **turbo factor**. Example: At a turbo factor of 7, the TurboSE sequence measures 7 times faster than a spin-echo sequence with comparable parameters.

The central echo, when the phase-encoding gradient is zero, determines the image contrast. The time interval between the 90-degree pulse and the central echo is the **effective echo time** TE_{eff}.

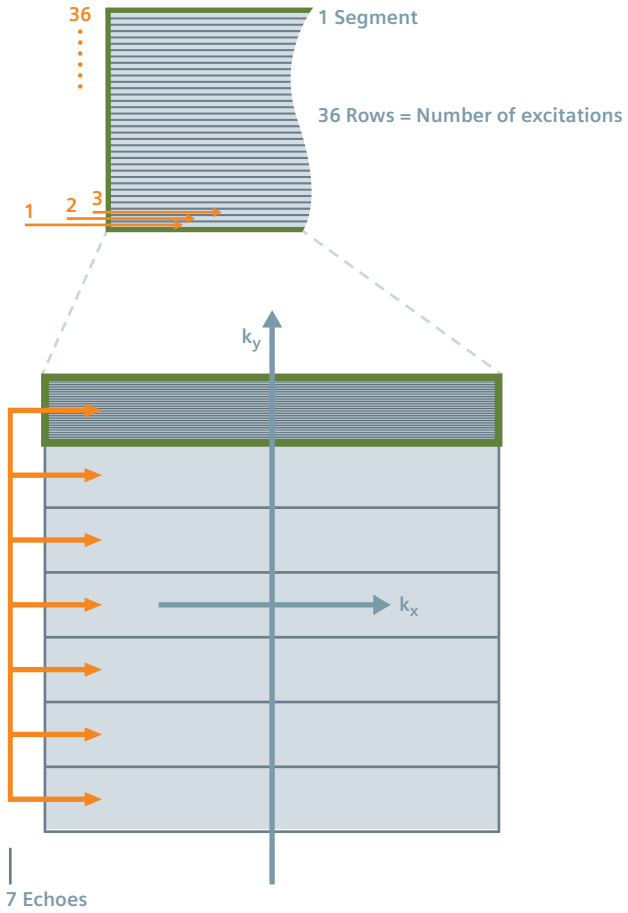


Segmented raw-data filling

TurboSE works because of the 'segmented k-space' technique: Within the repetition time TR, an entire series of raw-data rows is acquired, whereas with the conventional technique, only one raw-data row is acquired.

The raw-data matrix (k-space) consists of, for example, 7 segments (= turbo factor) of 36 rows each. The total number of rows is a whole-number multiple of the echo train length (7 × 36 = 252).

This means that within each echo train, one raw-data row is filled for each segment, creating a "comb" of filled-in rows. This "combing" has to be repeated 36 times, as shown by our example.



Optimizing parameters for TurboSE

The longer the echo train at a fixed TR, the shorter the measurement time. As a result, fewer slices can be acquired. At the same time, T_2 decay is stronger, which reduces the resolution in the phase-encoding direction, especially when tissue with a short T_2 is examined.

To ensure detection of small hemorrhages, for example, in the brain, a longer TR and a higher resolution are used for contrast improvement. The turbo factor may be reduced, for example, from 15 to five, but all in all this is still a significant increase in speed.

A further development of TurboSE techniques involves the combination with an inversion pulse (Turbo Inversion Recovery, TIR), with half-Fourier imaging (Half Fourier Acquired Single Shot Turbo Spin Echo, HASTE), or the addition of gradient echoes (Turbo Gradient Spin Echo, TurboGSE). Fastest turbo spin-echo techniques use a complete echo train of 256 or more echoes (single-shot TurboSE or RARE, Rapid Acquisition with Relaxation Enhancement).

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A TurboSE sequence accelerates image acquisition by gener-ating a series of spin echoes for each excitation known as the echo train.

The raw-data matrix (k-space) is segmented. If an echo train of, for example, 15 echoes (= turbo factor) is used, only 17 excitation pulses are required. The result is a significantly reduced acquisition time.

The contrast of TurboSE sequences is dominated by T_2-weighting.

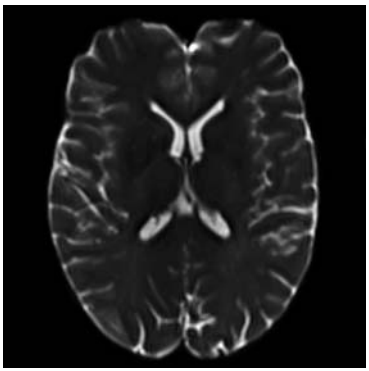
Echo-planar imaging (EPI)

Acquiring an entire image in one "shot"

From line scanning to plane scanning

In most of the conventional acquisition techniques discussed so far, like spin-echo and gradient-echo techniques, the raw-data matrix (k-space) is filled line by line with the generated echoes.

Echo-planar imaging (EPI) is a "single-shot" method. By that we mean that an EPI sequence uses a *single* excitation pulse to acquire an entire image, i.e. the whole 2D plane by generating an echo train of a sufficient number of gradient echoes to fill the raw data matrix.

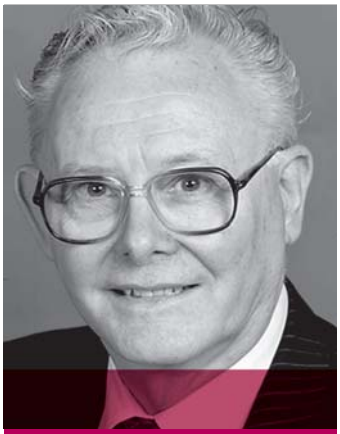


EPI image

An order of magnitude faster

EPI is one of the fastest techniques for acquiring MR images. The acquisition time for one slice is the repetition time, i.e. one TR period, a fraction of a second. The EPI sequence is repeated not for the phase-encoding steps, but for the number of slices chosen.

EPI is the method of choice for diffusion and cranial perfusion as well as for func-tional neuroimaging (BOLD imaging).



Echo-planar imaging was introduced in 1977 by Peter Mansfield (Nobel prize in 2003).

With EPI, "the speed of image formation can be increased by an order of magnitude or so over that of the single line-scanning method of imaging."

Sir Peter Mansfield

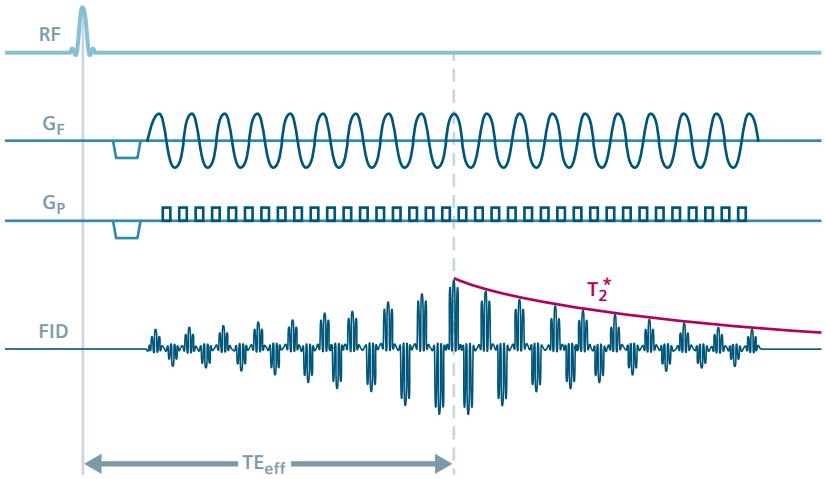
This is how EPI works

A readout gradient is switched in *bipolar* fashion. Within the FID, it generates an entire echo train of ascending and descending gradient echoes with alternating algebraic signs. The number of gradient echoes is the **EPI factor**.

The fast T_2^* decay of the FID leaves only approximately 100 ms to generate the echoes. For this reason, the readout is limited to between 64 and 256 echoes.

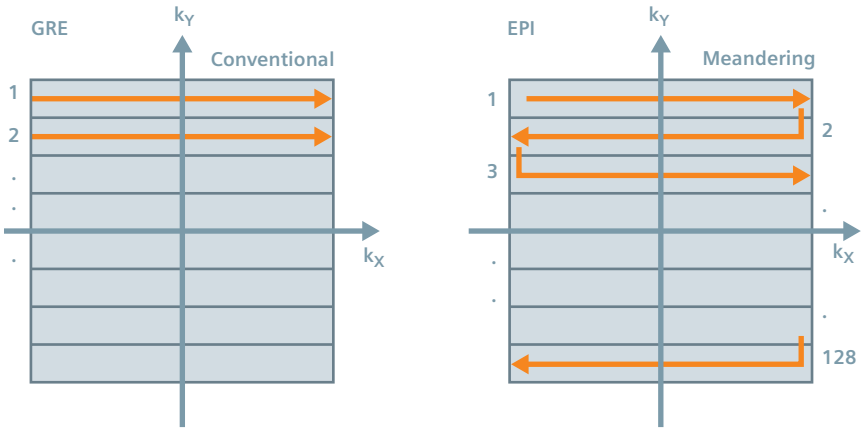
The EPI matrix is therefore between 64 × 64 and 256 × 256, accordingly the EPI factor is between 64 and 256.

The effective echo time TE_{eff} coincides with the maximum signal.



To move to the next raw-data row, the phase-encoding gradient is switched briefly between the individual gradient echoes (blips). The raw-data matrix is sampled in a "zig-zag" pattern, i.e. filled in a *meandering* manner.

This is how EPI sequences acquire diagnostic images in as little as 50 to 100 ms. These images are completely void of motion artifacts, making EPI especially suitable for examining dynamic processes or generating diffusion-weighted images, which are sensitive to motion on a molecular level.



Combining EPI for different contrasts

EPI is essentially a readout module, a kind of "add-on" to a pulse sequence. The EPI method can be combined with freely-selectable preparation pulses (spin echo, inversion recovery, etc.). This allows us to obtain different contrasts with EPI sequences. Since the echoes decay with T_2^*, the images contain a T_2^*-weighting component that varies with the basic contrast. As a single-shot procedure, EPI does not show any T_1 contrast.

EPI-FID sequences generate good T_2^* contrast that increases with the echo time.

EPI spin-echo sequences can be compared to conventional spin-echo sequences with an infinitely long TR. A long T_2 generates sharp images. For tissue with a short T_2, the image may not be as clear.

EPI diffusion sequences add additional diffusion gradients. They are sensitive to molecular motion and show the diffusion of water in tissue. EPI acquisitions have the advantage of freezing motion that would create artifacts in conventional sequences, obscuring the diffusion contrast.

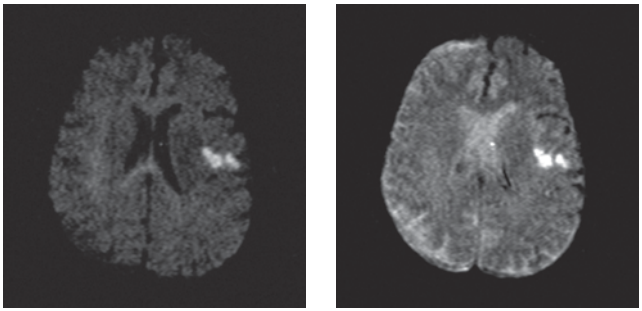


Image comparison: Strong diffusion contrast (left), weak diffusion contrast (right)

REDUCED TO THE ESSENTIALS

Echo-planar imaging is one of the fastest acquisition techniques. It uses a single exitation pulse to acquire an entire image within a fraction of a second: The whole plane in one “shot”. The complete echo train is created by a “blipped” gradient pulse. The EPI factor (for example, 128) defines the increase in speed.

The EPI method can be combined with preparation pulses, allowing for different contrasts (EPI SE, EPI diffusion, etc.).

A further development of Echo-planar imaging is segmented EPI, a multi-shot technique. The raw-data matrix is sampled segment by segment.

RESOLVE is a special sequence with segmentation in readout direction.

Going parallel

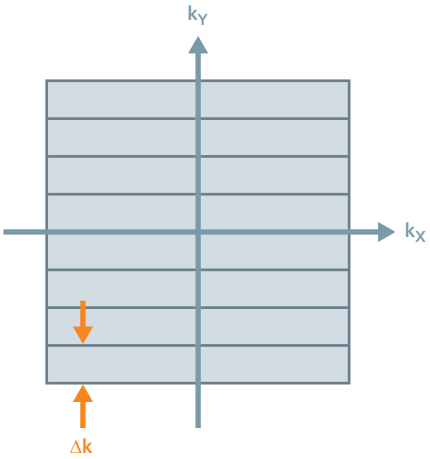
Parallel data acquisition using multiple coils

The limits of conventional phase encoding

In general, measurement time is proportional to the number of phase encoding steps. Standard fast pulse sequences acquire data sequentially: they fill the raw-data matrix (k-space) row by row (much like a fax machine). Each single row requires a separate application of gradient pulses. The maximum switching rates of the gradients are a limiting factor.

Example: To avoid motion artifacts, the patient has to hold breath approximately 20 seconds for each exposure involving a conventional cardiac examination. This can prove quite difficult for patients with serious heart conditions. The MR techniques introduced thus far have reached their limits. Even the fast EPI technique ist not suitable for all applications.

MR imaging can further be accelerated by parallel acquisition techniques (PAT) combining arrays of coils with dedicated postprocessing algorithms.

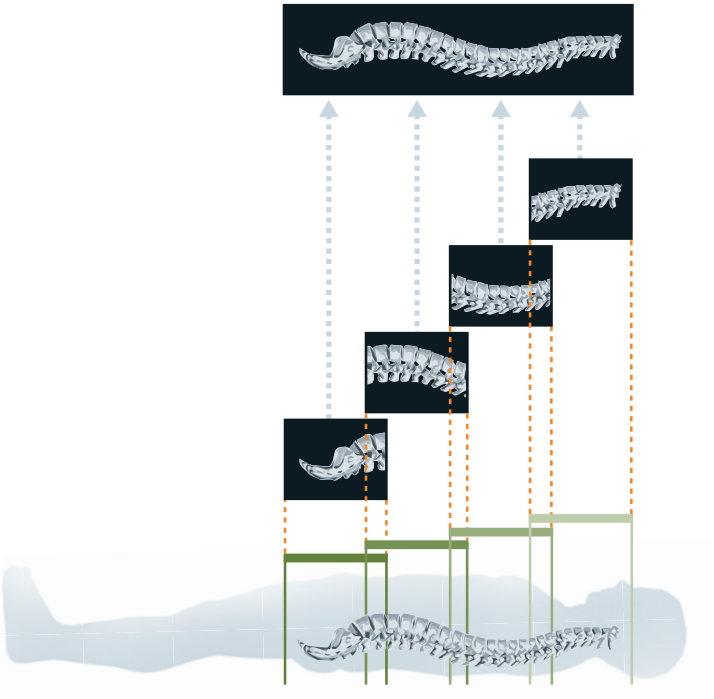


Arrays create the PAT factor

An RF coil is the receiver for MR signals. Let us assume we were to use not just one coil, but as many spatially arranged receivers as we would need for the resolution in the phase-encoding direction (somewhat similar to the methods used in a modern digital camera). In this case, we would not have to repeat a pulse sequence but could dispense with phase encoding entirely. This would considerably decrease our acquisition time.

Today's modern clinical methods involve parallel acquisition techniques that use several receivers simultaneously (from 4 up to 64). This configuration of several coil elements is known as an **array coi**l. Array coils werde previously used in *sequential* imaging.

In parallel acquisition techniques, the coil elements of the array are used to reduce the number of phase-encoding steps and ultimately the measurement time. The acceleration factor is known as the PAT factor.



The principle of array imaging: With a standard array technique, separate images are created for each coil element (in our example: 4). These images are subsequently combined into an overall image. In this way, we cover more of the body region to be examined without changing the measurement time.

Coil encoding supplements gradient encoding

Parallel acquisition techniques use the concept of the coil array. Unlike standard array techniques, they use the geometric characteristics of array coils.

The spatial arrangement of the individual coil elements provides additional information about the origin of the MR signals.

When the coil elements are arranged in the direction of phase encoding, we can utilize the additional information to omit some of the time-consuming phase-encoding steps. In other words, we supplement the spatial encoding via the gradients with encoding via the coils.

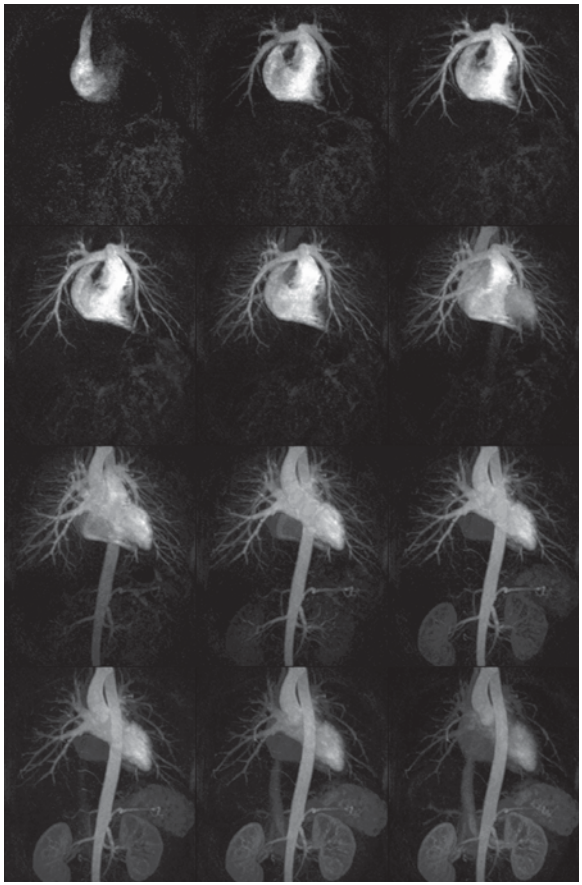
PAT allows either increased speed at the same image resolution or a higher resolution in the same acquisition time.

A shorter acquisition time when time is of essence is especially valuable (cardiac imaging in real-time, contrast-enhanced angiography, perfusion measurements).

Parallel imaging accelerated EPI sequences use shorter echo trains. The result is improved image quality as well as less smearing and distortions in the image.

Reconstruction methods for parallel imaging are SENSE, GRAPPA, or CAIPIRINHA (see Volume 2, *Magnets, Flow, and Artifacts*).

REDUCED TO THE ESSENTIALS



Dynamic MR angiography in parallel acquisition. Each individual 3D data set was measured in approximately two seconds.

Northwestern University, Chicago, Illinois

With parallel acquisition techniques, phase-encoding steps are skipped, which shortens the measurement time by the PAT factor.

Parallel imaging uses dedicated array coils and reconstruction algorithms (SENSE, GRAPPA, CAIPIRINHA).

In principle, this is the same as generating an image of a reduced field of view in conventional imaging.

Higher fields

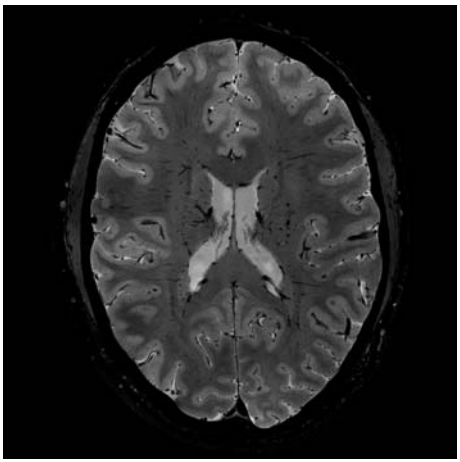
7 tesla and beyond

Impact of the magnetic field strength

At high field strengths, more information is available, enabled by high spatial and spectral resolution or by contrast mechanisms which are enhanced as field strength increases.

The primary reason to increase the field strength is to improve the MR signal. The magnetization M increases proportionally with the strength of the magnetic field.

The Larmor frequency is proportional to the field strength. At 7 tesla, proton spins will precess at a frequency of 300 MHz. The corresponding wavelength is as short as the dimensions of the human body. As we will see later, this has consequences for the design of the pulse sequences and the coil setup.



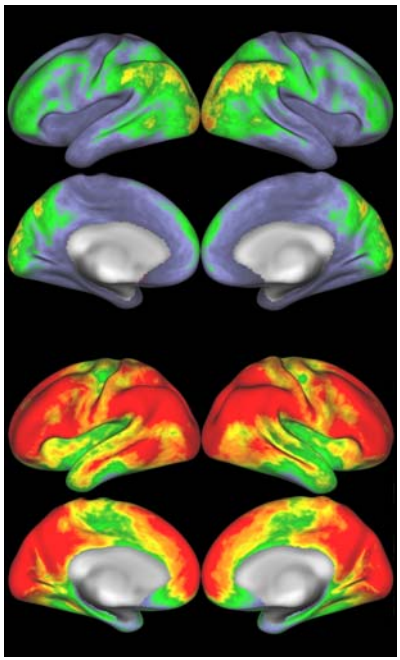
Ultra-fine anatomical details of the brain

Leibniz Institute for Neurobiology (LIN), OvGU, Magdeburg, Germany

Higher signal, better contrast

In MRI, the noise in images does not depend on the field strength. With higher field strength, the signal increases and the signal-to-noise ratio (SNR) is improved. Compared to 3 tesla, with a 7 tesla scanner we receive more than the double SNR which means a significantly improved image quality.

Some relaxation constants at 7T are dif-ferent to 3T. For some applications, this results in an increased contrast. BOLD imaging, for example, benefits from an increased SNR and an increased BOLD contrast simultaneously. For some applications, averaging is no longer necessary, and acquisition times are signifantly reduced.



Contrast-to-noise ratio maps in resting state MRI: low values (cold colors) for 3 tesla (top), higher values (warm colors) for 7 tesla (bottom)

Consortium
The Human Connectome Project. CMRR, Minnesota, USA; Washington University St.Louis, USA; Oxford University, UK

Increased resolution

The stronger MR signals of a more powerful scanner can be used to acquire images in the sub-millimeter range.

Let us assume we reduce the voxel size. As a result, the signal intensity decreases, since less protons are contributing to signal strength.But this is compensated by the higher field strength. At 7T, it is possible to acquire high-resolution morphologic MR images up to 0.2 mm isotropic voxel size.

Multichannel transmit and receive

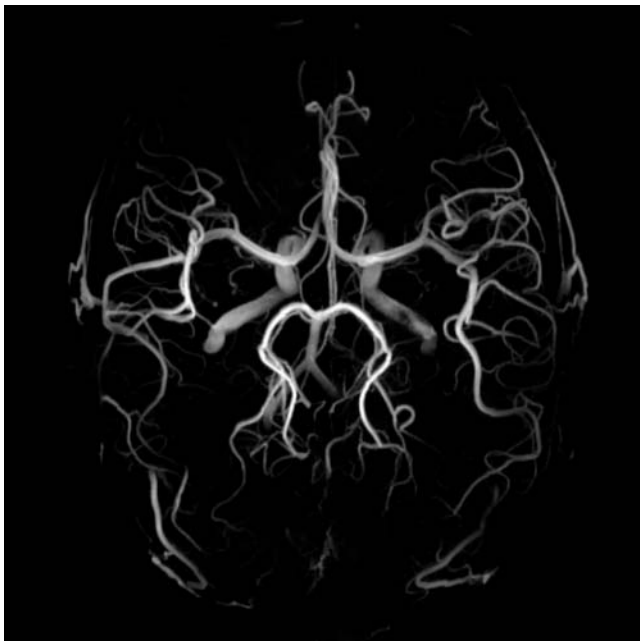
Challenges associated with higher fields include increased inhomogeneity of the RF field (B_1 inhomogeneity) and the static magnetic field (B_0 inhomogeneity). Additionally, at 7T, the RF energy absorbed by the human body, quantified by the specific absorption rate (SAR), is increased. Furthermore, other unwanted effects, such as susceptibility artifacts, scale with the field strength.

Increased B_1 inhomoheneity can be addressed by using multichannel parallel transmit (pTx). Due to the flexible exploitation of the degrees of freedom in the pulse design, the RF field can be homogenized over a predefined region and the SAR values can be significantly decreased.

Susceptibility effects can be addressed by using parallel imaging. At higher field strength, a higher acceleration factor can be achieved.

Brain angiography with 0.5 mm resolution

CMRR, Minnesota, USA



Clinical benefits of higher fields

High-resolution imaging allows to visualize anatomical details and functional information, for example, for analyzing tissue metabolism. This enables better lesion conspicuity. Some examples from many applications:

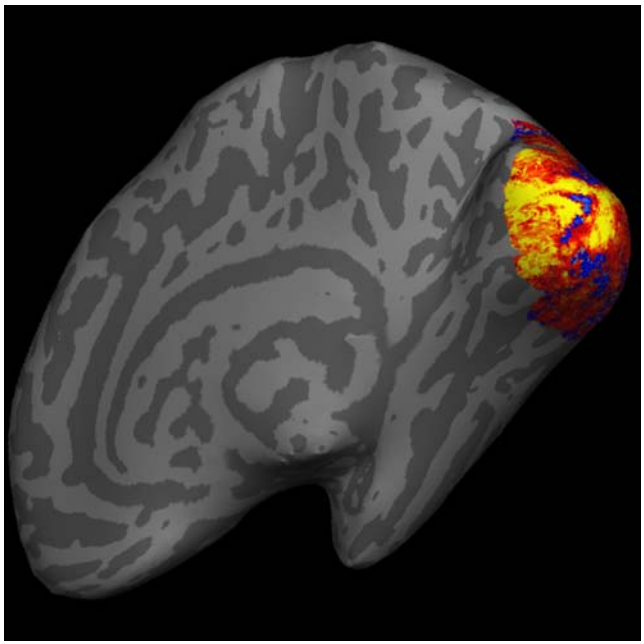
- Enhancing spectroscopy, functional neuroimaging (BOLD) or diffusion-weighted imaging
- Imaging without using an exogenous contrast agent at all. This benefits many SNR-limited applications in the body, such as angiography or perfusion.
- Imaging of low-sensitivity nuclei other than hydrogen

Transparent fibers showing crossings all over the brain: diffusion-weighted EPI with 1 mm resolution

Max Planck Institute, Leipzig, Germany

High resolution BOLD fMRI

MGH, Boston, USA



Hybrid imaging

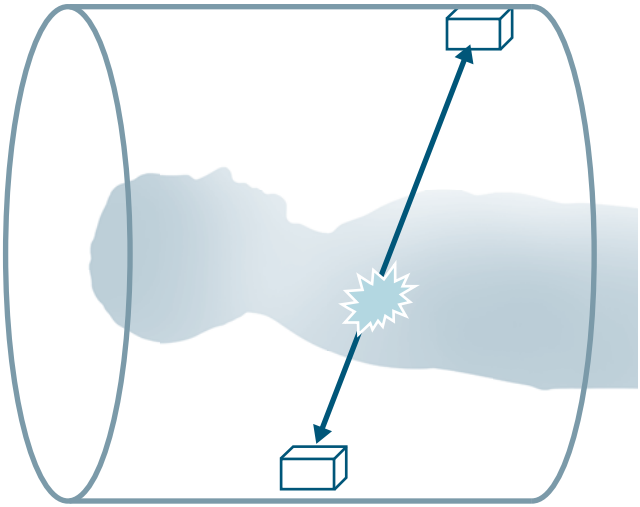
Combining different imaging modalities (MR-PET)

What is PET?

PET (Positron Emission Tomography) is an imaging technique for localizing tissue with high metabolic activity which is often an indication of tumors. The tissue can be indirectly localized through metabolite accumulation of a radioactive tracer.

As the tracer decays, it emits a positron (the antiparticle of an electron). The positron loses energy through interaction with surrounding tissue. Depending on the endpoint energy of the tracer or isotope utilized, the positron will travel a certain distance until it annihilates with an electron. Subsequently, two gamma rays, consisting of high-energy photons, are emitted in nearly opposite directions. If a detector pair registers both back-to-back photons within a certain time window, a 'line of response' (LOR) is formed between both events and registered as a 'coincidence'.

From a large number of coincidence events, three-dimensional images of the tracer distribution in the body can be calculated.



Detection of photon emission

MR-PET advantages

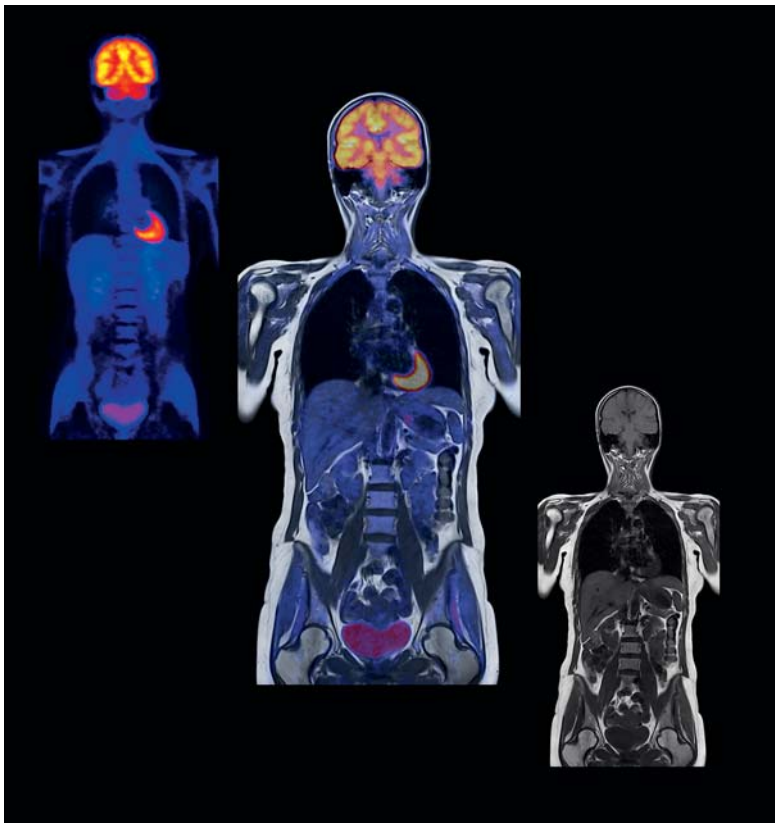
Let us compare PET with MRI. As the generation of PET signals is based on events of positron annihilation, the resulting images are comparatively noisy and of lower spatial resolution.

However, PET provides metabolic details which cannot be generated with less sensitive MR imaging. Thanks to better localization and higher resolution, MR brings structural and additional functional information to the image.

In this respect, the combination of PET and MR is highly complementary. Despite the exposure to ionizing radiation, no lack of diagnostic information occurs.

PET (on the left), fused, and MR image (on the right)

ZEMODI, Bremen, Germany



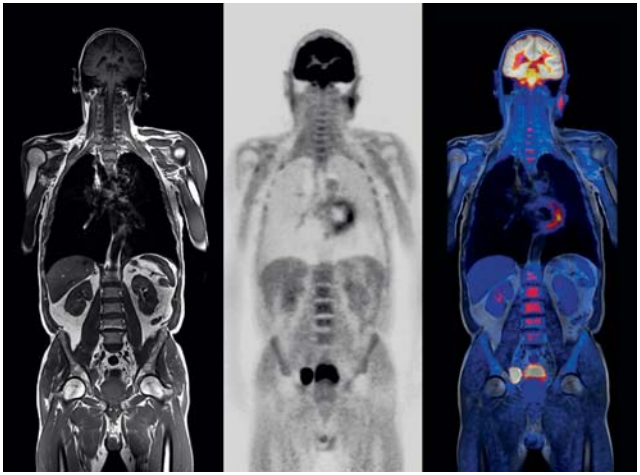
MR-PET team play

How do these modalities work together?

MR can affect PET performance, due to the high magnetic and radiofrequency fields. MR image quality, however, can be impaired by either radiofrequency noise introduced by the PET electronics or magnetic field inhomogeneities.

One approach is sequential acquisition: PET and MR images are processed separately and aligned afterwards. In this case, there is no mutual interference. However, sequential imaging holds the risk of involuntary patient motion in between the two examinations and, therefore, may increase the chance of local misalignment. In addition, the examination is time-consuming.

This disadvantages are compensated for by acquiring PET and MR images simultaneously during identical (functional) states.



Whole-body images: MR (on the left), attenuation corrected PET, fused MR/PET (on the right)

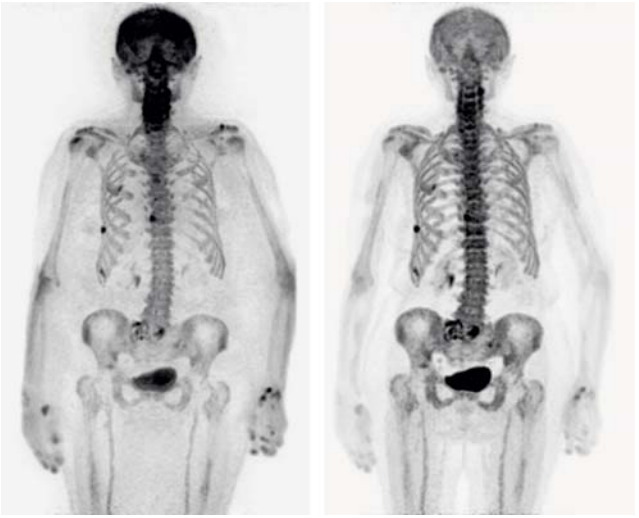
ZEMODI, Bremen, Germany

PET signal attenuation

The signal of emitted photons is weakened by material or tissue. For example, RF coils, the patient table, and the patient (soft tissues, air, or bones) weaken the PET signal.

Therefore, acquired PET data must be corrected for artifact-free PET imaging. This is done with specifically post-processed MR images known as **attenuation correction** (AC) maps. These AC maps are used to create corrected PET images.

While attenuation-corrected images are generally more reliable, the correction process itself leads to artifacts. As a result, both corrected and uncorrected images are always reconstructed together. For appropriate diagnosis, the uncorrected images have to be checked as well.



Whole-body PET images: uncorrected (on the left), attenuation corrected (on the right)

ZEMODI, Bremen, Germany

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