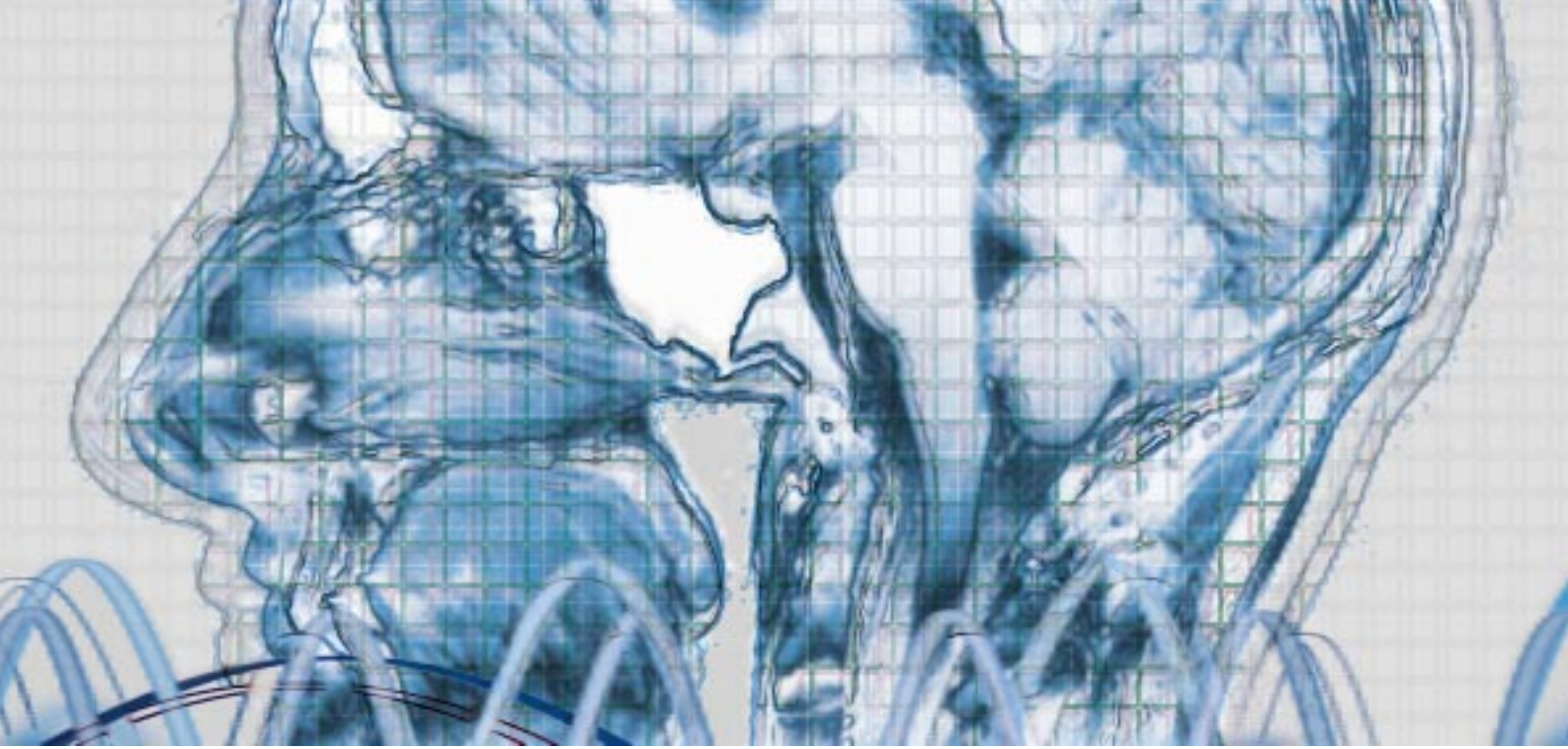


Magnets, Spins, and Resonances

An introduction to the basics of
Magnetic Resonance

SIEMENS
medical

Magnets, Spins, and Resonances

Magnets, Spins, and Resonances

An introduction to the basics of Magnetic Resonance

Siemens Medical Solutions
Magnetic Resonance
Erlangen

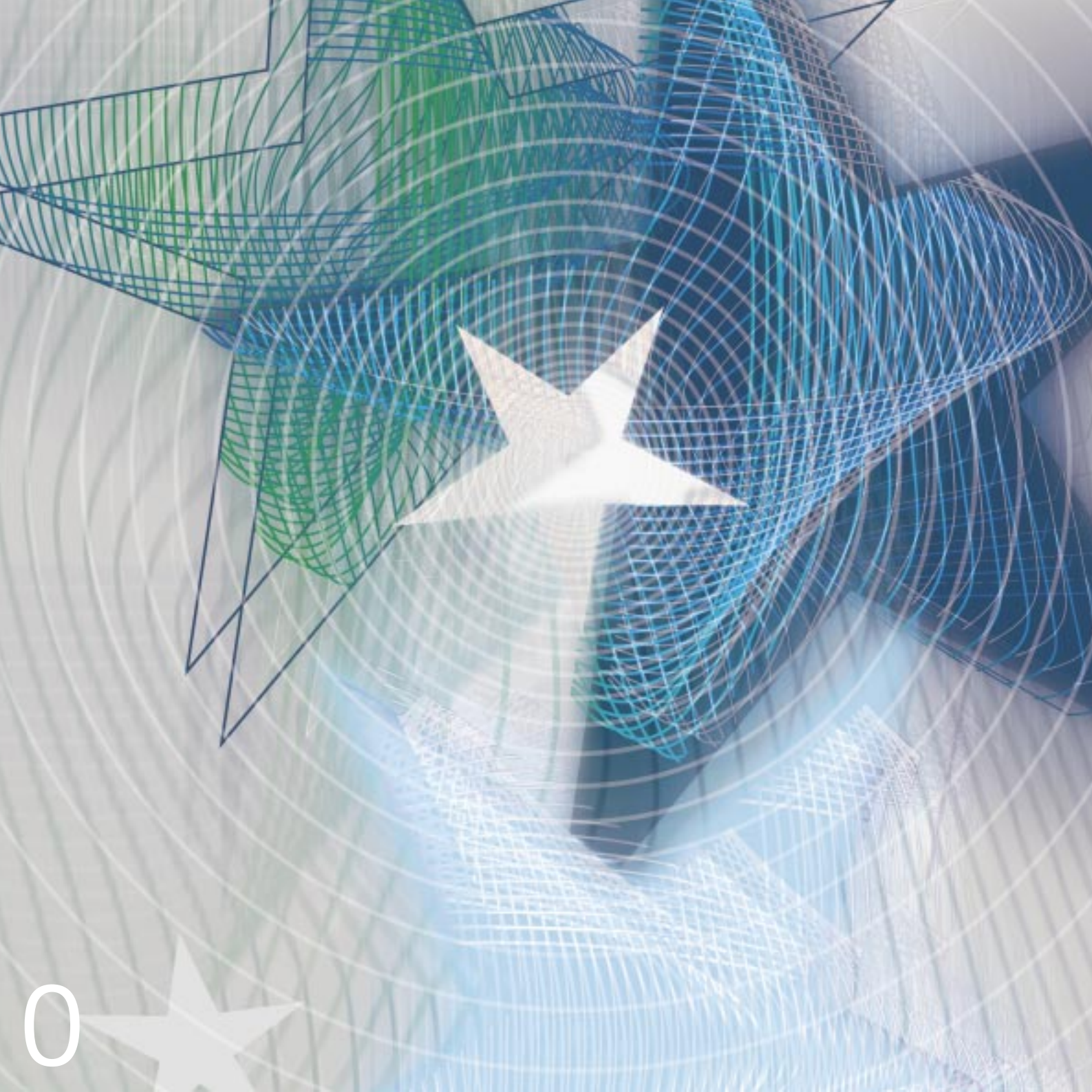
Magnets, Spins, and Resonances

Join us in our journey through the fascinating world of modern MR imaging.

This brochure and its scope of application address a wide audience, covering topics of interest to radiologists, radiological technicians, medical specialists, as well as researchers. Additionally, the brochure is also an excellent entry-level source of information to those with an abiding interest in Magnetic Resonance.

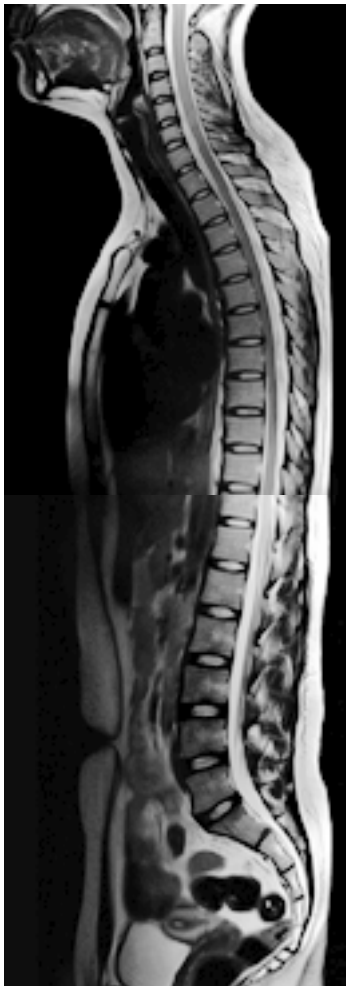
In this vein, we hope you are enjoying the brochure as highly informative, easy-to-understand reading material.

Siemens Medical Solutions



MR Highlights

Morphology—details from head to toe



MR is a non-invasive imaging technique. Its primary field of application includes the display of morphology, that is, tissue structures in a series of slice images through the body.

The advantages of MR imaging

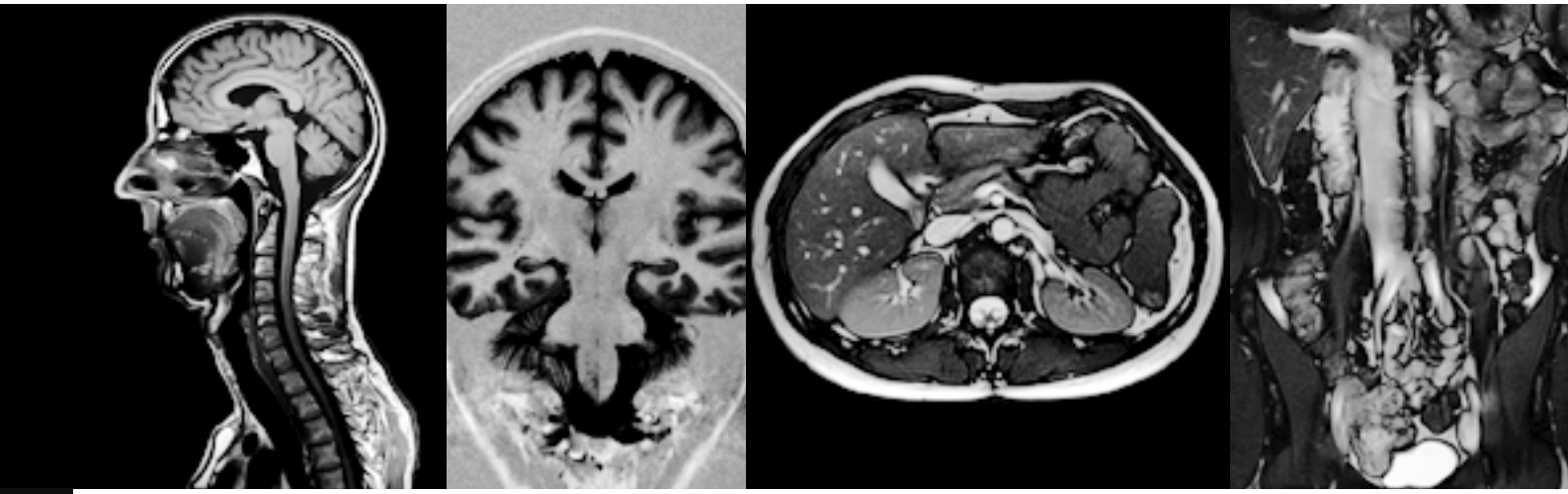
The three main advantages of MR imaging are:

- excellent soft tissue contrast with high resolution
- display of several images and oblique cuts
- no ionizing radiation

Modern MR systems allow for fast images of the body from head to toe. For example, examinations of the entire spine can be completed in as little as two steps.

As pacemaker of the MR industry, Siemens has incorporated a unique coil concept known as Integrated Panoramic Array (IPA™).

Combined with automatic table positioning (Integrated Panoramic Positioning—IPP™), MAGNETOM systems by Siemens allow for the quick (high-speed) display of large volumes.

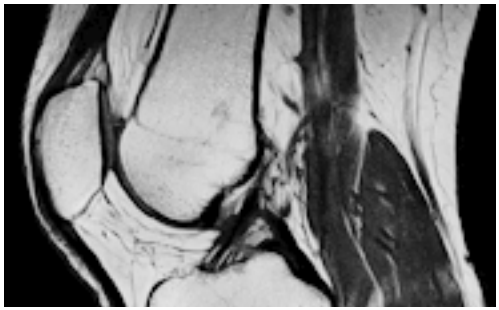


MR-imaging provides image contrasts that result from the combination of several parameters. These include

- the density of the nuclear spins stimulated, especially that of hydrogen protons
- the relaxation time for magnetization of the tissue under examination, as well as
- various different contrast mechanisms

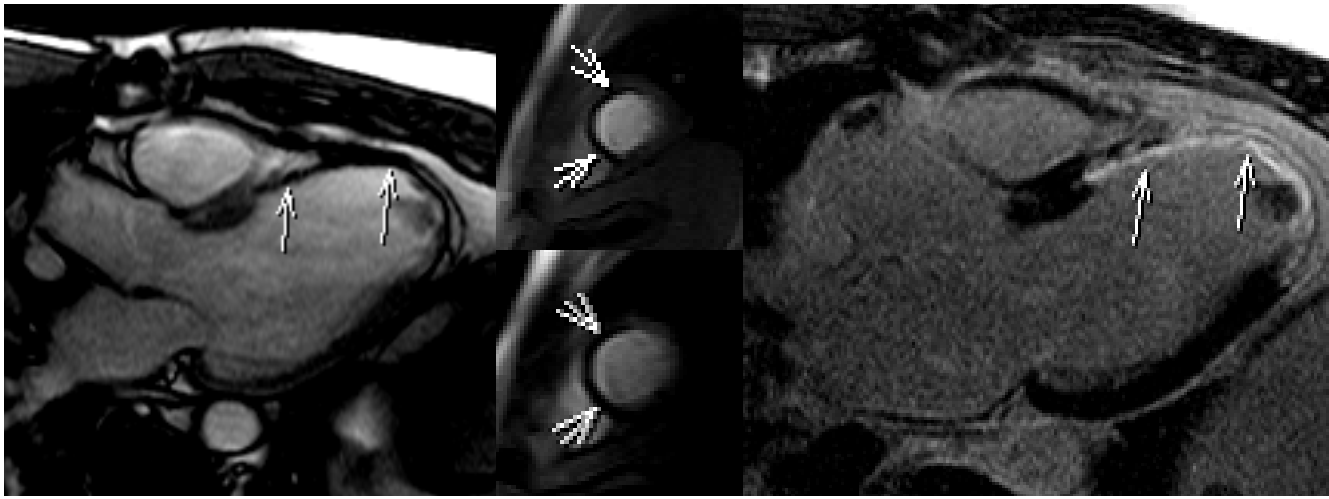
The different MR contrasts enable precise diagnosis while supporting tissue characteristics.

High-resolution MR images with a small field of view show excellent anatomic details.



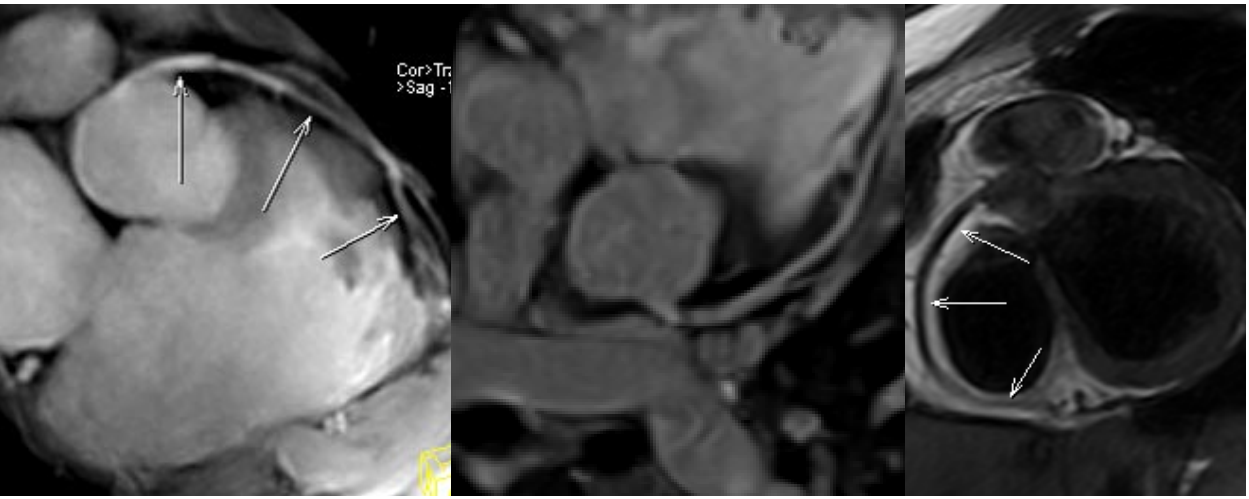
Comprehensive imaging of the heart

Cardio-vascular MR imaging (CMR) profits greatly from the ability of magnetic resonance to display slice images in any orientation with high spatial as well as temporal resolution. The prerequisites for a diagnostically useful image are performance-oriented gradients, superb pulse sequences as well as a robust, high-speed hardware.



MR imaging of the heart provides excellent morphological displays.

It also provides a multitude of information regarding the function of the myocardium, such as vitality, ejection fraction, wall movement or valve functions.

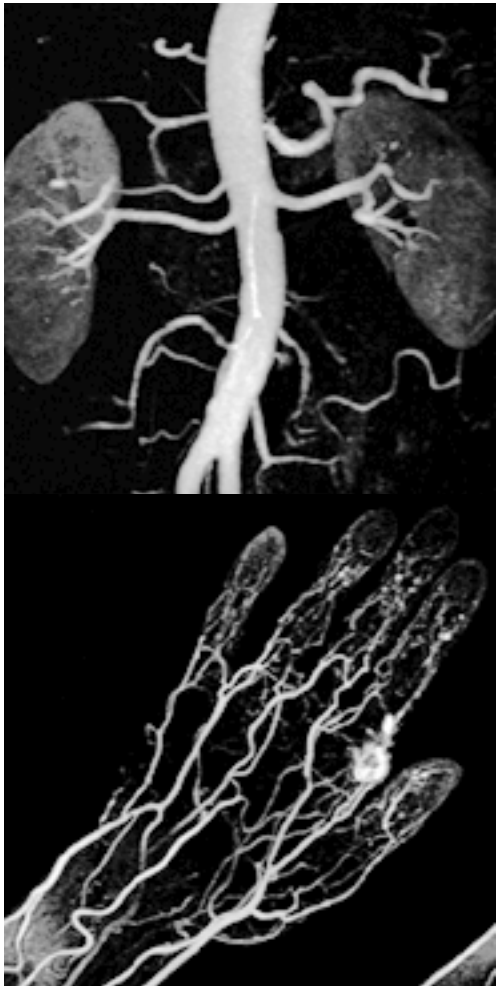


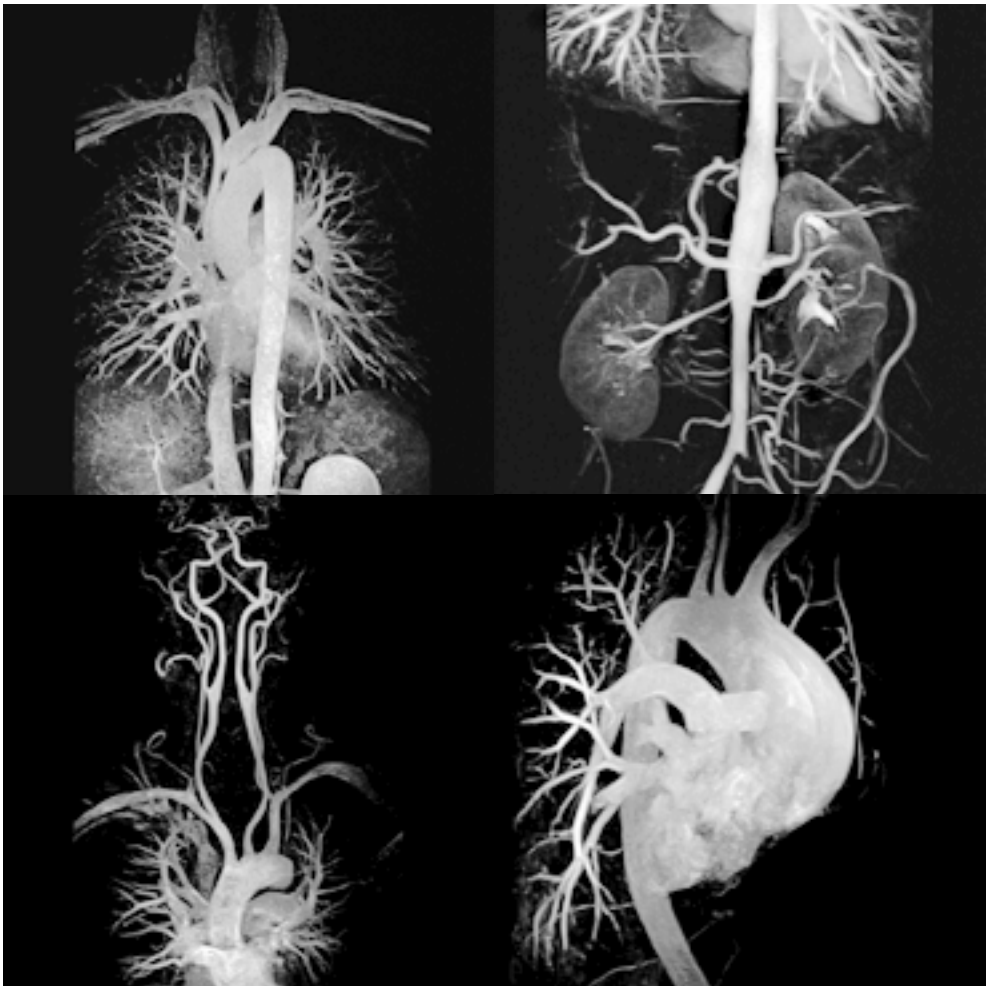
MR-imaging offers methods using contrast agent to display coronary vessels. To visualize coronary arteries, however, MR provides for methods that do not use contrast agents. These are the so-called TrueFISP and Dark Blood techniques.

Contrast-enhanced angiography from head to toe

Considerable advances have been made in the area of contrast-enhanced MR angiography.

The interaction of strong gradients, high-speed MR systems and Care Bolus results in excellent contrast at optimal contrast agent consumption.





Contrast-enhanced MR angiography using strong gradients, iPAT (integrated Parallel Acquisition Techniques) and array coils.

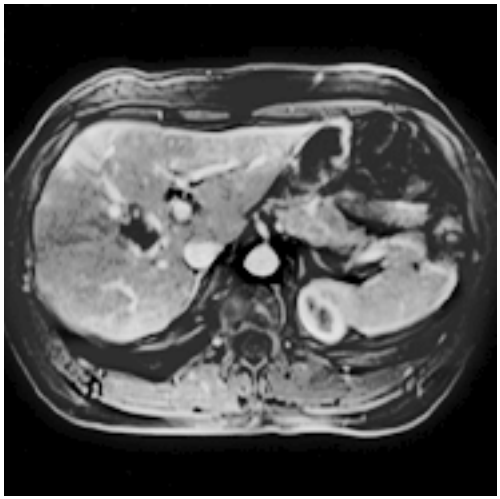
Excellent detail recognition of blood vessels in a matter of seconds.

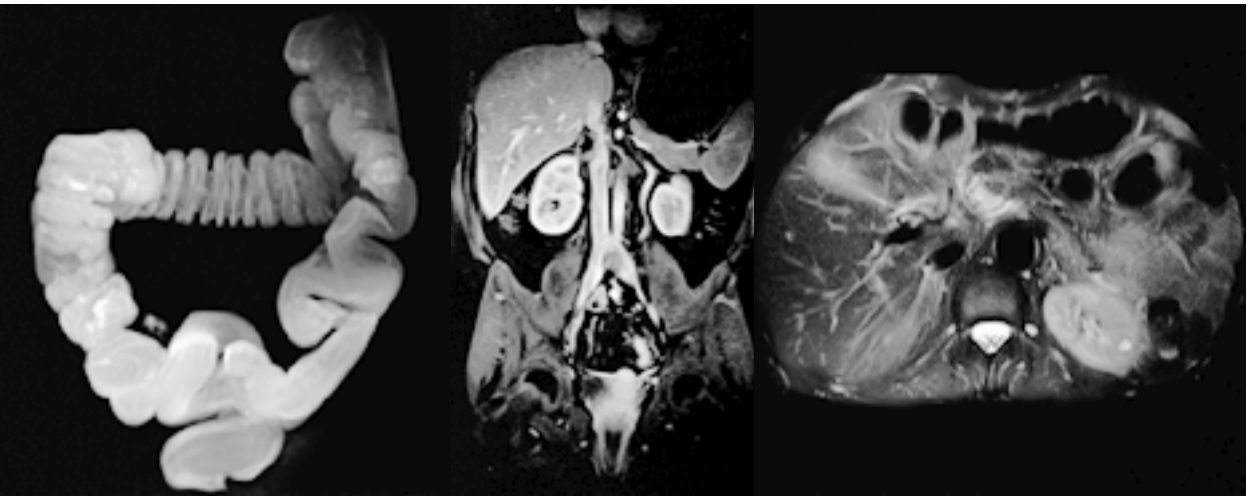
Gastroenterology and MR

MR imaging is a highly suitable technique for gastroenterology.

New as well as unique pulse sequences by Siemens, such as 3-D VIBE (Volume Interpolated Breathhold Exam), enable the display of anatomical details as well as dynamic angio information.

3-D VIBE with *fecal tagging* finds extensive use in MR colonography.





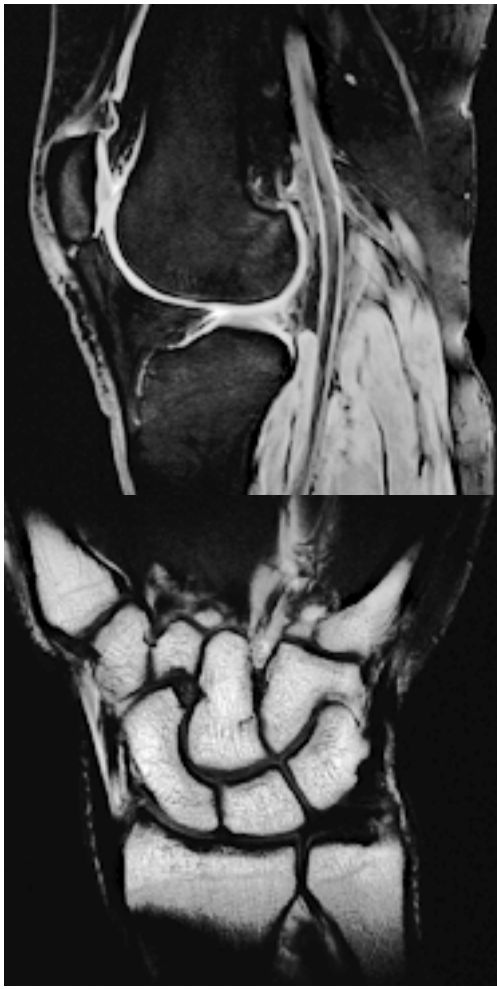
New techniques such as iPAT (integrated Parallel Acquisition Techniques) or PACE (Prospective Acquisition CorrEction) facilitate examinations and help reduce motion artifacts.

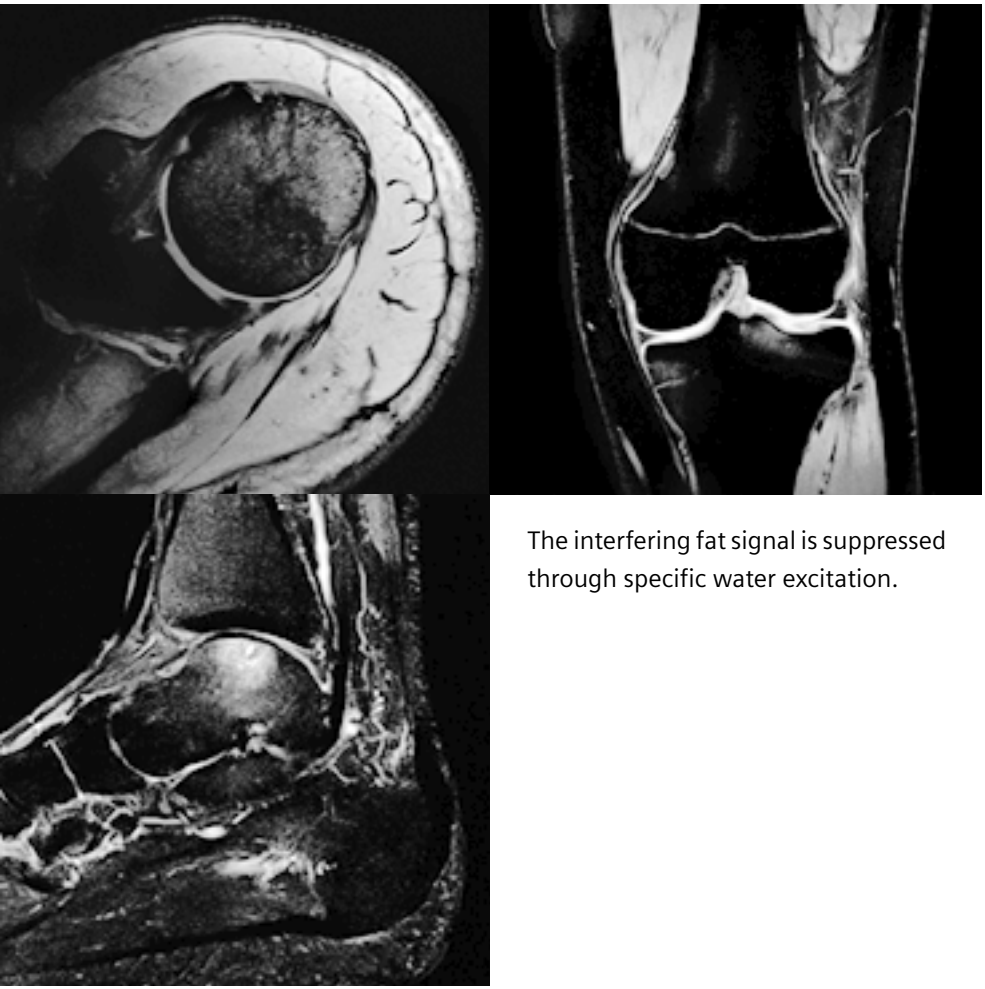
New views in virtual endoscopy are obtained by post-processing 3-D data sets.

Orthopedics in MR

High-resolution imaging of joints and interarticular spaces

High-resolution images with good contrast are the basis for satisfactory diagnosis. The images are generated with unique pulse techniques, such as 3-D DESS (Double Echo Steady State) and MEDIC (Multi Echo Data Image Combination).





The interfering fat signal is suppressed through specific water excitation.

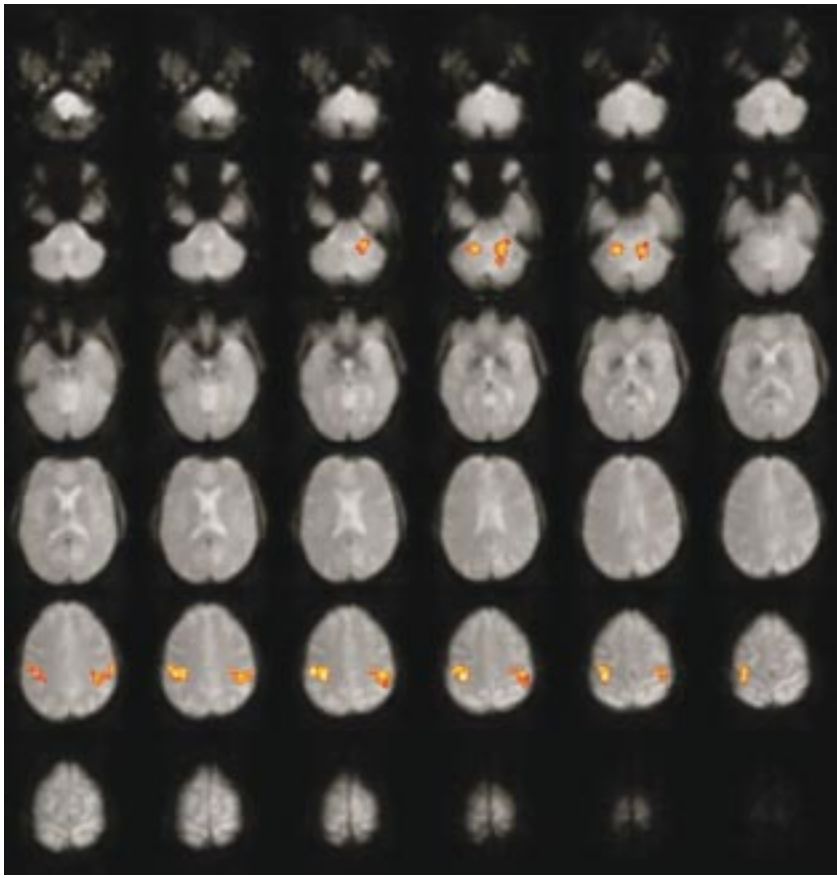
Neurology and comprehensive, high-speed diagnosis with MR

Neuro-imaging presents one of the most revolutionary applications of magnetic resonance.

Inline technology enables automatic computation and superimposing t-test (z-score) images on anatomical EPI images.

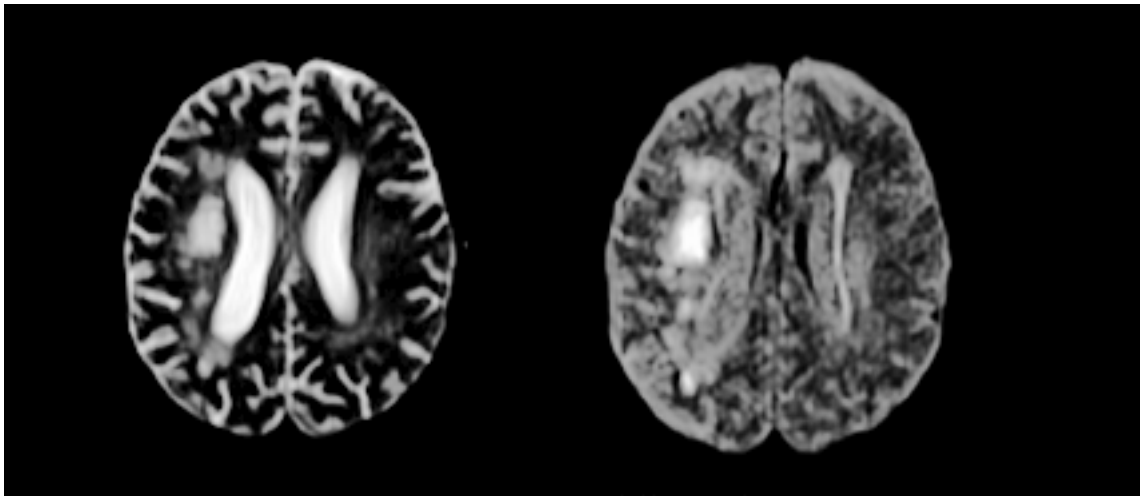
ART (fully automatic motion correction) as well as spatial filtration support accurate results.

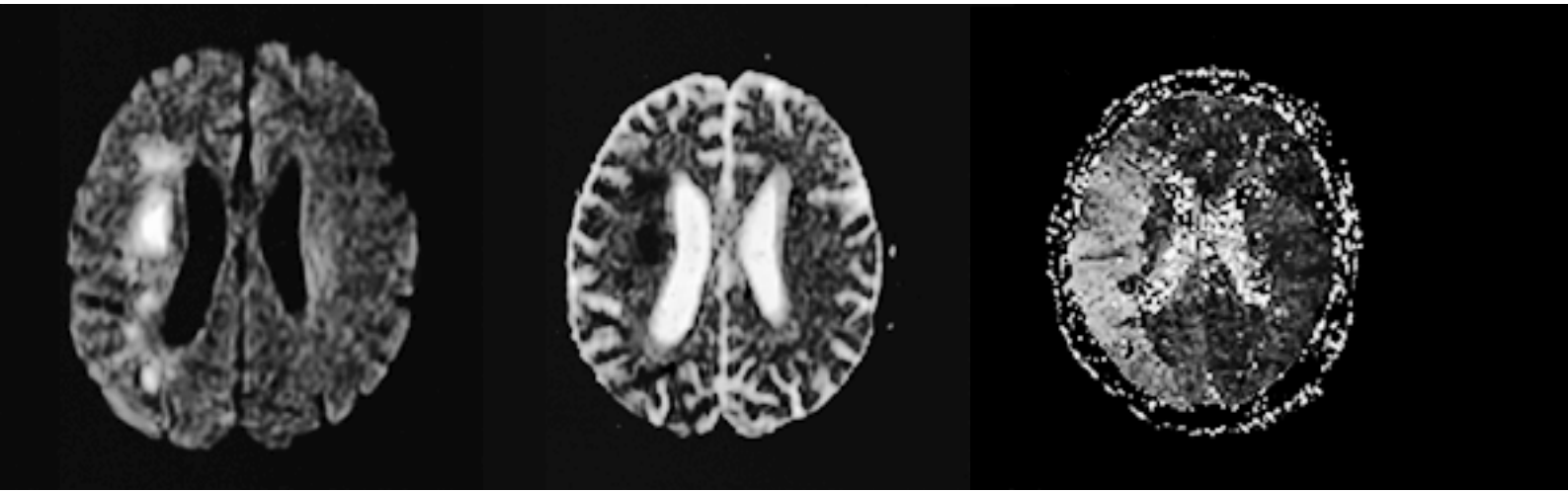
Modern technology enables the compact display of mosaic images, very helpful, for example, in OR planning.



Diffusion and perfusion imaging

Diffusion imaging with Single-Shot-EPI sequences provides 16 different b-values with a maximum b-value of 10,000 s/mm^2. The ADC cards (Apparent Diffusion Coefficient) as well as the trace-weighted images are automatically computed via integrated post-processing (inline).

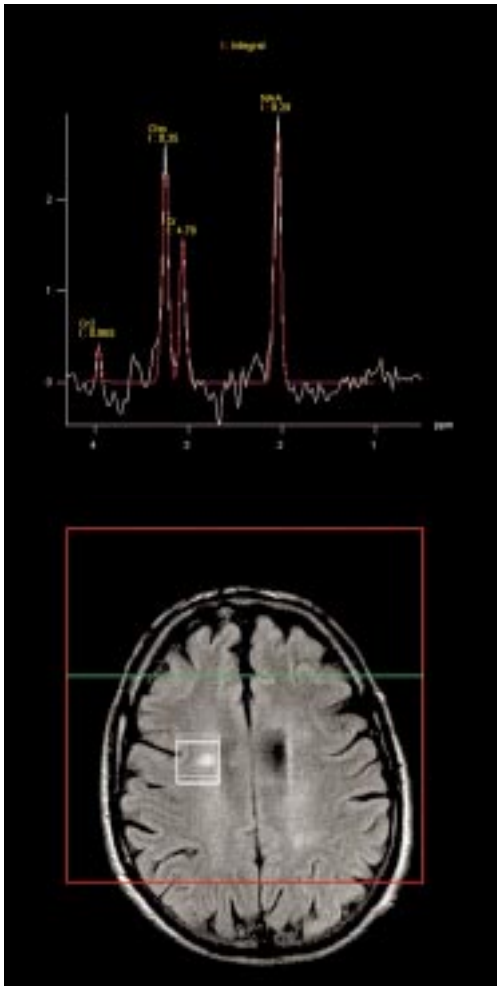




Perfusion imaging with inline computation of the Global Bolus Plot (GBP), the time-to-peak map (TTP) as well as the percentage-of-baseline-at-peak (PBP). Inline computation greatly accelerates neurological examinations.

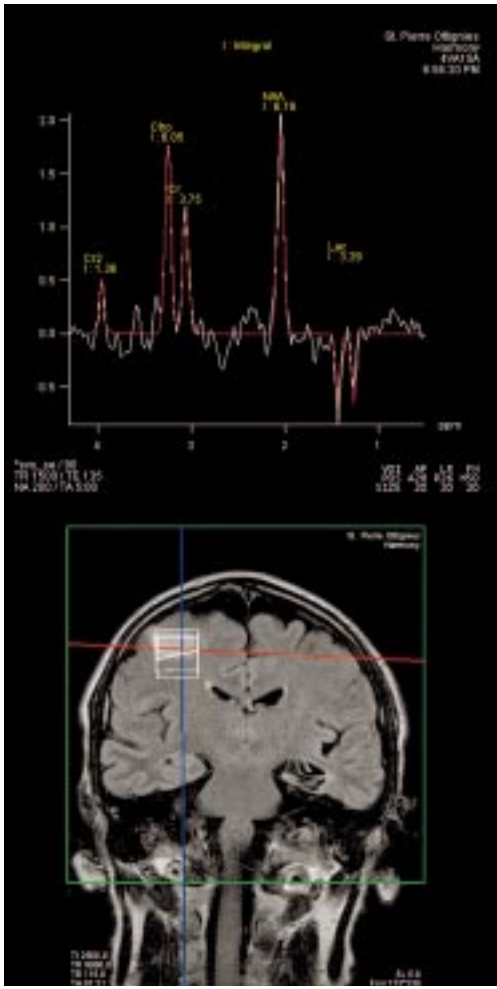
Proton spectroscopy

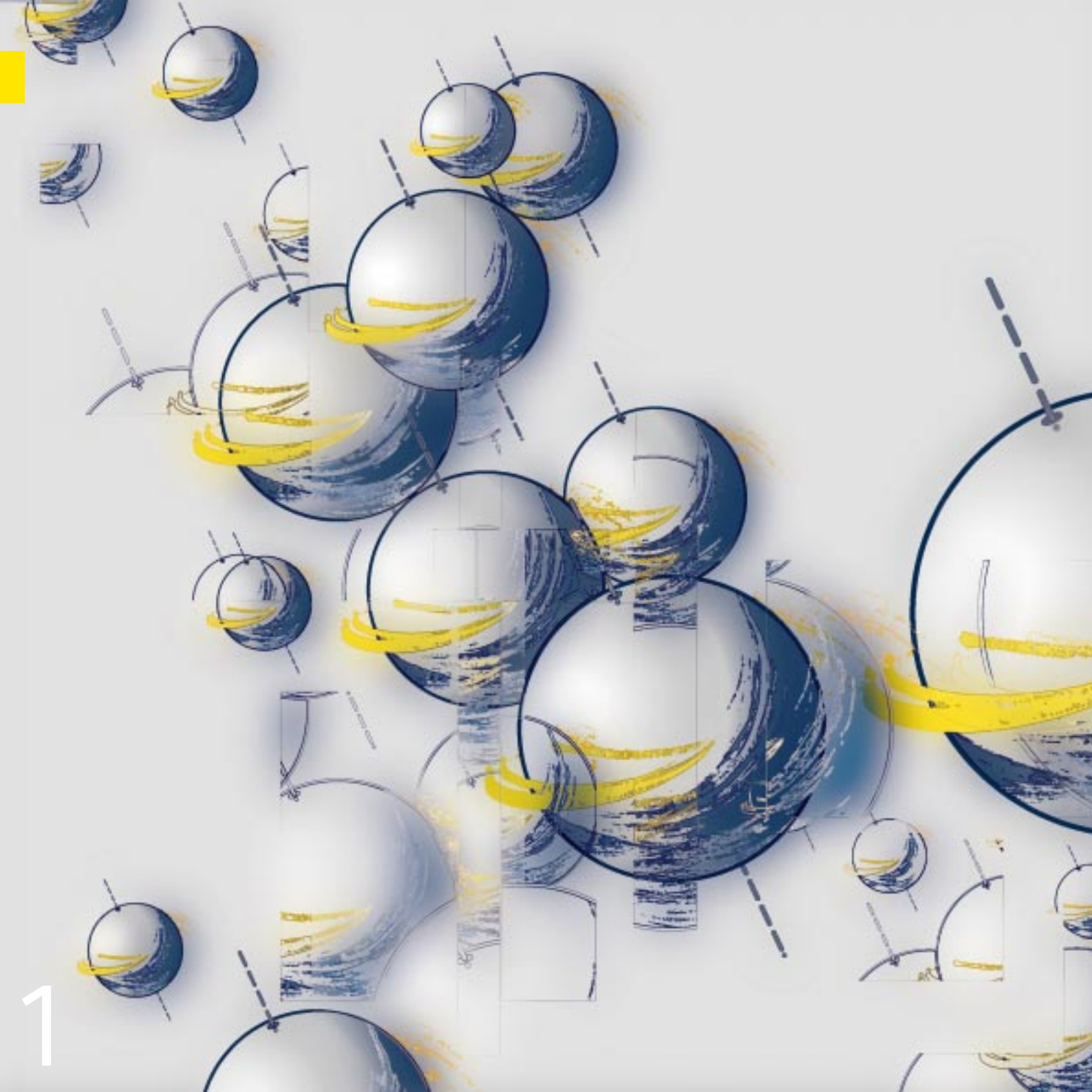
In addition to image generation, MR spectroscopy provides for biochemical quantification.



Over the years, clinical MR spectroscopy has changed from rather complex into simple procedures.

Modern spectroscopy uses new pulse sequences with shorter echo times. The new evaluation software provides for color metabolite images as well as spectral overview cards.





A short excursion through MR Physics

What does an MR examination involve?

Let's follow a patient examination step-by-step. The first steps involve moving the patient into the magnet where he is exposed to a strong magnetic field. During the course of the examination, the patient's body generates magnetic reactions that produce a measurable signal.

To sufficiently explain these reactions, we would like to invite you to join us on our short excursion through MR physics. During this trip, it will become quite obvious that the MAGNETIC SPIN is so-to-speak the power behind MAGNETIC RESONANCE or MR TOMOGRAPHY.

Nuclei and spins

Magnetic resonance or MR tomography. Our focus will be on the magnetic spin and its magnetic effects. That is why we start our trip by looking at the atomic nuclei present in the human body. True, beginnings are never easy. So it's best to simplify matters as much as possible.

Hydrogen is the least complex

Atoms of chemical elements consists of a nucleus and its electronic shell. Hydrogen is the most prevalent element and possess the least complex nucleus: in this case a single, positively charged PROTON.

MR tomography uses the *magnetic* characteristics of the hydrogen proton to generate images.
The two advantages that hydrogen brings to MR tomography are:
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 give the *strongest* magnetic resonance signal.

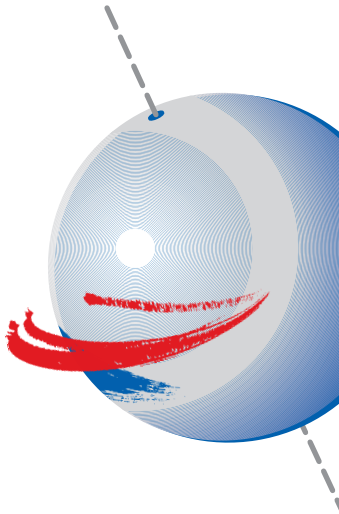
Protons and billiard balls

What makes the hydrogen protons useful for MR?

Protons possess a singular characteristic: the spin.

The SPIN is a purely → **quantum-mechanical** characteristic of atomic building blocks, but it is best to imagine that we could actually "see" a rotating proton, just like

- the spin of a billiard ball
- the rotation of the earth about its axis
- the spinning of a child's top.



The special features of a spin

While ordinary objects can rotate at different speeds, the spin of a nucleus *always* remains the *same*; it is a *unique* property of the nucleus. The only variation you will see is the change in the direction of the axis. And there is one more difference with respect to a billiard ball: the spin *never* rests.

Why are we focusing on the spin?

The spin is the source of the magnetic resonance signal: a nucleus with spin is always *magnetic*.

FOR DISCUSSION

Classic physics or quantum physics

Our model that likens the spin to the "rotation" of a sphere is nothing more than an analogy. You cannot apply it to all atomic nuclei or all shapes of spin.

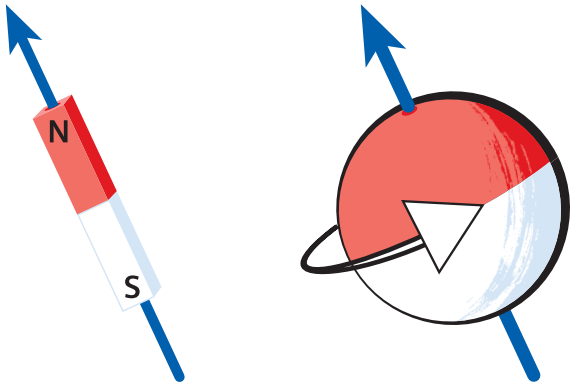
But the following does apply, independent of any kind of analogies: the spin is a measure of the *quantum state* of an atomic particle. It can be accurately defined through complex state vectors. MR tomography can be applied without in-depth knowledge of quantum physics.

Note that MR imaging is not described by individual spins, but rather by their collective characteristics. Fortunately, this leads to easy-to-understand models that can be used with undue distortion of reality.

Bar magnet and spin magnets

Although the characteristic of spin is purely quantum mechanical, this does not keep us from giving it a simple model—in this case a bar magnet. This type of magnet has a magnetic *north pole N* and a magnetic *south pole S*.

Let's assume that the proton acts like a tiny bar magnet, although we will see below that this is somewhat misleading (and will be discussed later).



Because, in classical physics, a spinning charge generates a magnetic field, it is easy to conclude that the magnetism of the proton is caused by the → **rotation of this charged particle**. We call this magnetic force the SPIN MAGNET.

Spins are characterized by direction

Although individual spins may point in various directions, we can view their effects as a single → **vector**, an aligned magnitude in space. The randomly selected direction of the spin magnet runs from the magnetic south pole to the north pole (as shown by the blue arrow).

Of course, it is not the proton itself that is a vector, but rather its spin and/or magnetic effect.

In what follows we are not going to look at the protons themselves. Instead we are going to study their coupled characteristics: spin and magnetism. This is what we mean when we are talking about a "spin magnet".

FOR DISCUSSION

The rotating charge

Classic physics consider the electrical CHARGE of the proton as the source for its magnetic effect: the moving charge is nothing more than an electrical current. In turn, this current generates the associated magnetic field. A rotating charge usually generates its magnetic effect in the direction of the rotating axis. This magnetic force is known as the MAGNETIC MOMENT.

As compared to a proton, the electrically neutral NEUTRON does *not* have a charge. But since it still has a spin, it is considered useful for magnetic resonance.

This means that an electrical charge is not a prerequisite for the magnetism of a nucleus. Actually, the modern theory of quarks postulates the reverse effect, namely that magnetism is the cause for an electrical charge.

What really matters is the right direction

The essentials about vectors and arrows

Do you feel like going over VECTORS one more time? There are a large number of physical magnitudes, for example, temperature or mass, etc. that are known to be non-directional. What this means is that they are sufficiently identified by magnitude and units (e.g. 21 degrees Celsius, 5 kilograms).

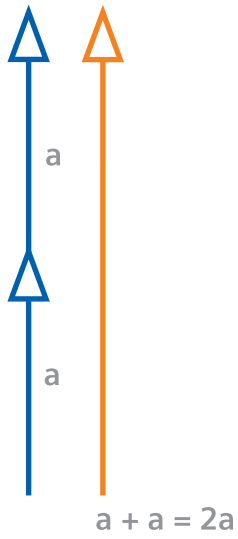
Spin magnetism, however, is a directional magnitude. But the magnitude of magnetism alone does not tell us about its effect. To determine that we need to know its direction.

Again, there are a multitude of physical magnitudes that depend on spatial orientation (e.g. force or speed). Vectors are a suitable means for depicting these magnitudes.

ARROWS are excellent for depicting vectors. The direction of the arrow corresponds to the orientation of the vector quantity, the length of the arrow corresponds to the magnitude of the vector.

Vector quantities allow for SPATIAL ADDITIONS. The direction has to be taken into account and visualized by linking the arrows.

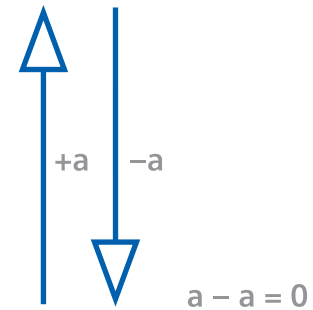
If the arrows point into the same direction, the magnitude of the vector sum is simply the sum of the magnitudes: (in this case $\boldsymbol{a}+\boldsymbol{a}$*).*

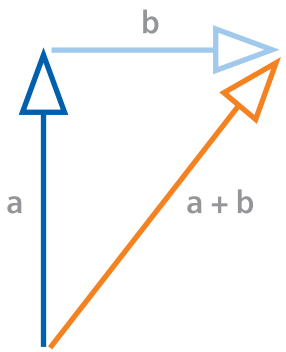


Vectors of the same magnitude CANCEL each other: ***a*** *−* ***a*** *=* ***0***

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 the COORDINATE SYSTEM.

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





Please do not confuse vectors with arrows and vice versa. A vector is a mathematical model for a physical quantity. An arrow is merely a tool for the visual display of a vector.

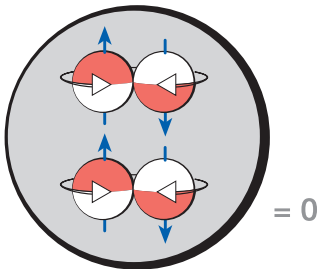
Which nuclei are suitable for use with magnetic resonance?

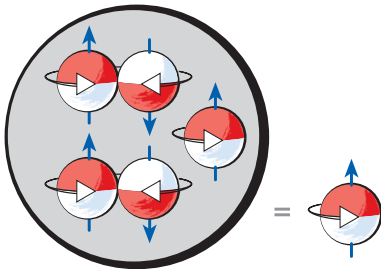
We would now like to continue with the atomic nuclei of other elements.

Protons and neutrons are ATOMIC PARTICLES. Both have the property of spin.

Atomic 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}. Two thirds of the isotopes found in nature have a net nuclear spin, making them suitable for use with magnetic resonance.





Atomic 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 with magnetic resonance.

Review

The nuclear spin is the source of the magnetic resonance signal: a nucleus with spin is always magnetic.

The spin is a directional quantity. Spins, just as vectors, allow for spatial addition.

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

FOR DISCUSSION

How do we get a nuclear spin?

In the atomic nucleus, two identical particles cannot be in the same state. They have to align their spin orientation *anti-parallel* to each other, and the net spin of this "couple" of particles cancels. This means that the "dancing couple" is invisible to the outside. This rule of nature is known as the PAULI EXCLUSION PRINCIPLE. It's the "solo dancers" that create the nuclear spin.

The nuclear spin, as a quantity resulting from individual spins, is not a rotation of the atomic nucleus as such. In the strictest sense, this applies to the individual proton as well. Its spin results from its internal structure (quarks and gluons).

How magnetization is created

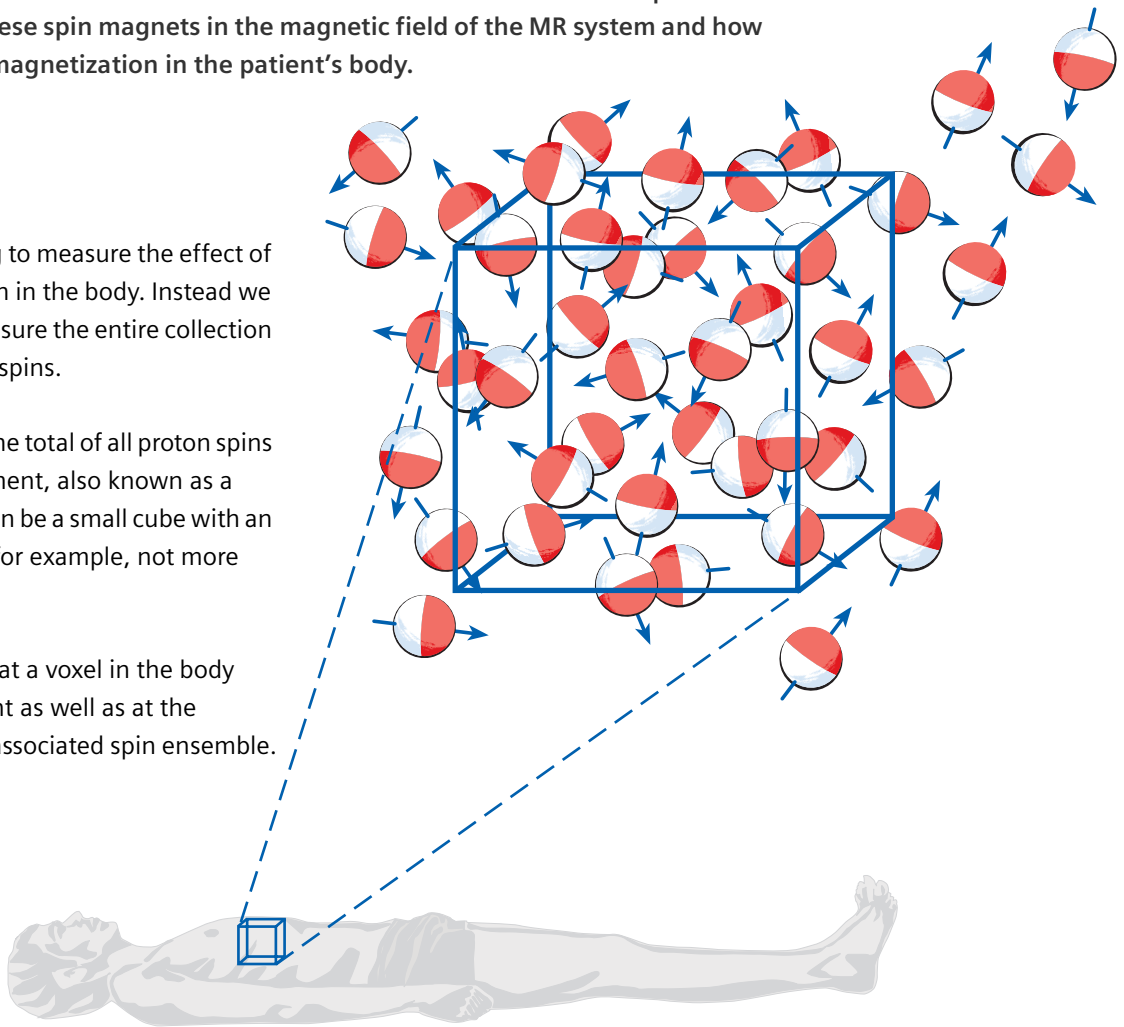
Protons and atom nuclei with a nuclear spin can be simplified by visualizing them as spin magnets. How do we benefit from this model? It allows us to explain the alignment of these spin magnets in the magnetic field of the MR system and how they generate magnetization in the patient's body.

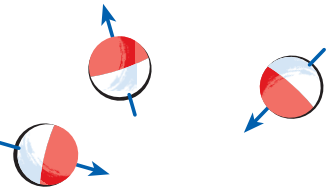
Voxels and spin

We are not going to measure the effect of an individual spin in the body. Instead we are going to measure the entire collection ("ensemble") of spins.

An ENSEMBLE is the total of all proton spins in a volume element, also known as a VOXEL. A voxel can be a small cube with an edge length of, for example, not more than 1 mm.

Let's take a look at a voxel in the body tissue of a patient as well as at the behavior of the associated spin ensemble.





The spin ensemble in field-free space

The effect of the ensemble is created by the spatial addition of the individual spin vectors.

Without an external magnetic field, in → **field-free space,** the spins in the ensemble are randomly oriented and their effects cancel each other. This is the reason why the ensemble appears to be *non-magnetic*.

FOR DISCUSSION

Can we actually talk about a field-free space?

A completely random orientation of spins applies only to an absolutely field-free space. The protons "feel" the magnetic field of the earth at all times.
While the magnetic field of the earth is approximately 20,000 times weaker than that of an MR magnet, it is nevertheless effective. In other words, even though the magnetic field of the earth is very weak, the ensemble is magnetically affected by it outside the MR system.

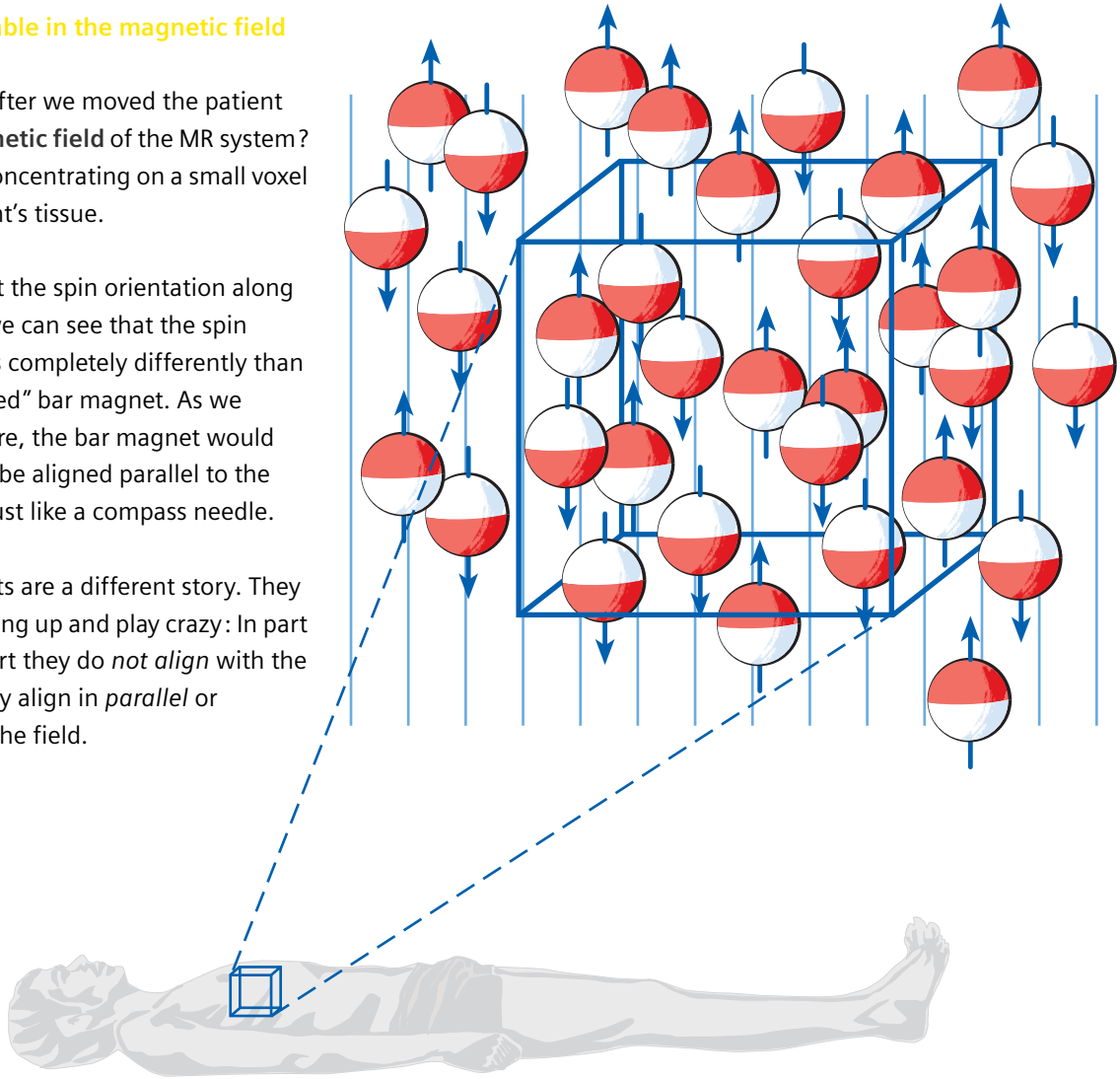
For this reason, magnetic resonance can be used in the magnetic field of the earth (e.g. for the discovery of subterranean oil fields). However, for clinical imaging, magnetic fields with ten thousand times this strength are a must. The patient is therefore positioned in the strong magnetic field of the MR magnet.

The spin ensemble in the magnetic field

What happens after we moved the patient into the → **magnetic field** of the MR system? Let's continue concentrating on a small voxel inside the patient's tissue.

When we look at the spin orientation along the field lines, we can see that the spin magnet behaves completely differently than our "well-behaved" bar magnet. As we mentioned before, the bar magnet would obey nicely and be aligned parallel to the magnetic field just like a compass needle.

The spin magnets are a different story. They certainly are acting up and play crazy: In part they *align*, in part they do *not align* with the field, that is, they align in *parallel* or *anti-parallel* to the field.

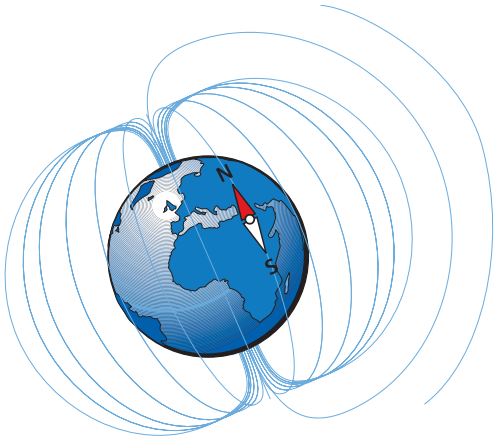


What you should know about magnetic fields

Each magnetic field exerts a force on magnetic and magnetizable particles, including our spin magnets. The effect of this force is symbolized by magnetic FIELD LINES.

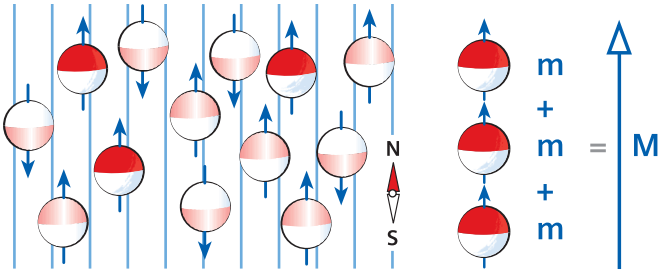
The strength of this force at each location in space is known as the magnetic induction. In MR technology, the term MAGNETIC FIELD STRENGTH at units of 1 tesla is commonly used. 1 tesla is approximately 20,000 times stronger than the magnetic field of the earth.

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.



Magnetization—generated by the excess spins

In tissue the static magnetic field generates a preferred direction of the spins parallel as well as anti-parallel to the field lines: → **Spin up** and **spin down** are the two preferred spin orientations in the magnetic field.



The ratio of up and down spins is *not* 50:50, otherwise the spins could cancel for the most part. Instead there is a small majority of EXCESS SPINS, the up spins to be precise. This leaves the down spins in the minority.

The excess spin magnets (**m**) add up to a macroscopic effect, known as the MAGNETIZATION (**M**) of the ensemble.

This magnetization is *very weak* (paramagnetism) as compared to the well-know magnetism of iron (ferro magnetism).

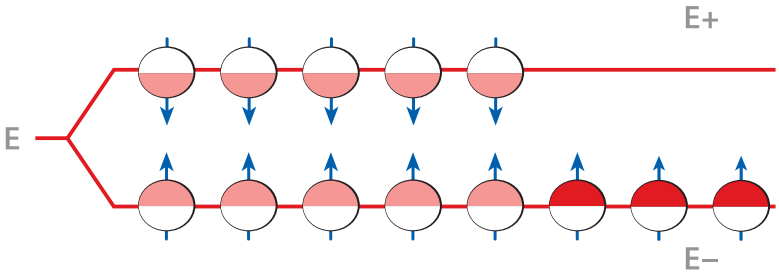
In passing, we would like to mention the following: the magnetic field *not only* affects hydrogen protons, it affects *all* nuclei with spin as well as electrons. For simplicity's sake we are only focusing on hydrogen protons relevant for MR imaging.

FOR DISCUSSION

Spin up—Spin down

The source for magnetization of the ensemble is the energy split of the spins in the magnetic field. The two spin orientations, up and down, correspond to two different states of energy. The UP SPIN has a lower energy (E–) than in the field-free space (E); the DOWN SPIN has a higher energy (E+).

The lower energy state is the preferred in the magnetic field: *more* spins jump into the lower energy state (E–) than into the higher one (E+). It takes a certain time for the magnetization to regrow. After its recovery, the ensemble is in a state of *energy equilibrium*.



At this point, you may be asking yourself: if there are more spins with lower energy in the magnetic field, the total energy of the spin ensemble must have *dropped* as well.

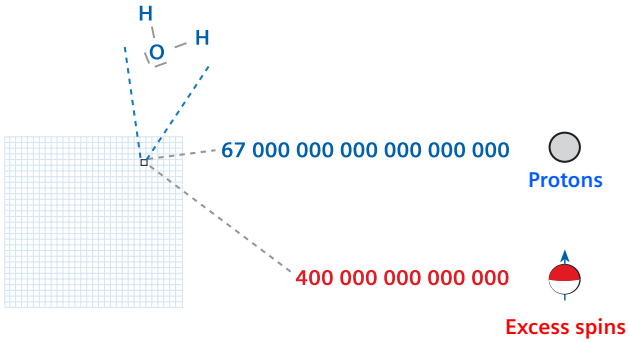
You are quite right. Protons are *not alone* in space. They are surrounded by what is known as a LATTICE. As magnetization regrows, the protons release energy to the lattice. The spin ensemble cools down just like a warm spoon that has been immersed in cold water.

This "cooling down" is the reason underlying magnetization of the spin ensemble in a magnetic field.

The energy equilibrium between the two levels is truly *dynamic*: the spins jump in pairs from up to down and vice versa (they literally flip-flop). The ratio between up and down spins remains the same—and as a result the magnetization to the outside.

Calculating the excess

At this point we know that: the slight majority of up spins is known as excess which results in the magnetization of an ensemble in the direction of the field. How many excess spins do we actually have?



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

- grows in proportion to the PROTON DENSITY
- grows with the strength of the external magnetic field, however
- drops as the temperature increases.

At body temperature and a field strength of 1 tesla, there are approximately 6 excess spins, or 0.0006 % of 1 million protons.

Since it takes many zeros after the period to express it in percentages, there is another way to express this 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: our voxel with an edge length of 1 mm holds 1 cubic millimeter of water, or 1 micro liter. This volume contains approximately $6.7 \cdot 10^{19}$ hydrogen protons. At 1 tesla this translates into approximately 6 ppm excess spins. This means that 400 trillion small spin magnets **m** add up to a macroscopic magnetization **M**.

Review

When the human body is exposed to a strong magnetic field, it will acquire a slight magnetization along the field lines.

The source are the nuclear spins in the tissue. The spins align with the magnetic field, however, at random distribution.

Most nuclear spins cancel each other. It is the total sum of excess spins that creates magnetization to the outside.

FOR DISCUSSION

Highly discrete, yet continuous

We are not quite finished with the model for our up and down spins. While it explains the generation of magnetization along the field lines, it does not explain how the spins produce an MR signal. The model needs further refinement.

True, we may simplified matters regarding the magnetic field. The individual spins are not at all—even though it is a belief held by some—strictly aligned in either the up or down state. For quantum mechanical reasons, the protons take on a *superimposition* of the two spin states (a spin jumps into the up or down state only during a measurement).

Let's compare what we just said with a manual car transmission. While driving the car, you shift gears, and the speed of the car changes continuously. The same is true for the spin of a proton: it has two discrete *basic states*, up and down, but it may also be in the transverse plane.

Spin precession in the magnetic field

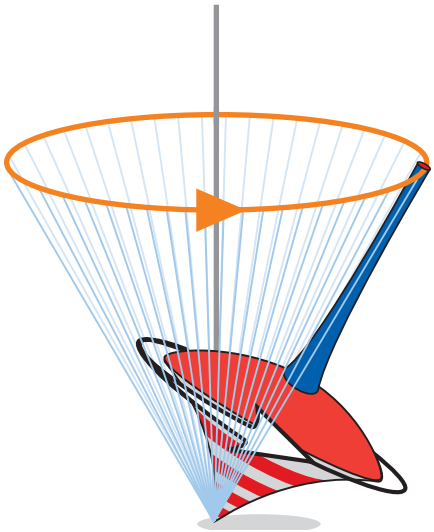
Spin magnets behave completely differently from bar magnets that would line up perfectly in one direction in the magnetic field. While spin magnets do not line up like compass needles, they do possess a unique characteristic that allows for magnetic resonance: they precess.

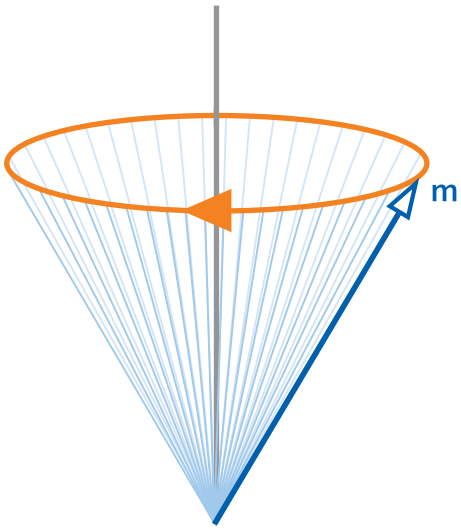
Spinning tops

As children, we all liked tops and 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 moves: its axis of rotation moves in the shape of a cone about the direction of gravity.

This kind of movement is know as PRECESSION.

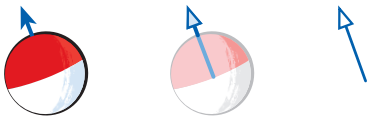




Magnetic tops in the body

And this is the movement of a spin: Just like a top, a spin subject to a magnetic field has to move in the shape of a cone in the direction of the field. The spin magnet behaves just like a magnetic top. We call this the SPIN PRECESSION.

Note that the proton itself is *not* spinning like a top. It is rather its spin and or spin magnet (**m**). To emphasize this difference, we are letting the ball disappear for good...



Radio frequencies in the magnetic field

The speed or characteristic → **frequency** of a spin "rotating" about the external field direction is of great importance in magnetic resonance. It depends on

- the type of nucleus and
- the strength of the magnetic field applied.

The stronger the magnetic field, the faster the spin. The precession frequency of a 1 tesla magnetic field is twice that of a 0.5 tesla field.

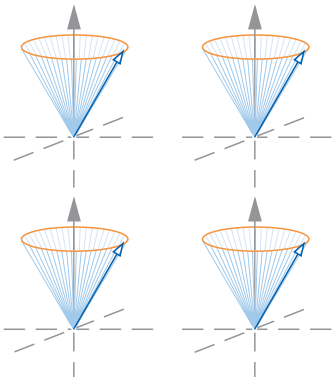
The precession frequency is also known as the LARMOR FREQUENCY.

How important is the Larmor frequency for magnetic resonance?

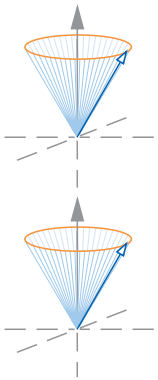
The highlights:

Just like radio signals, signals from a group of precessing spins can be received once the necessary technical requirements are in place.

For this purpose, the technology of the MR system has to be tuned to the Larmor frequency of the spins. You can compare this to turning the button on the radio to receive a different station.



Peaks and crests

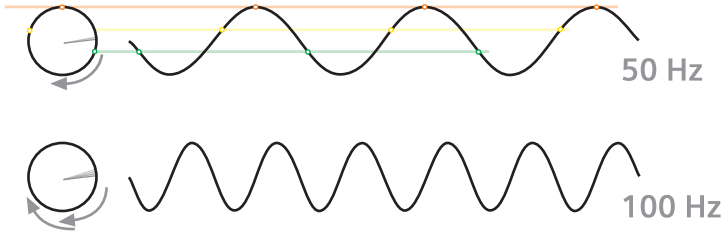


About frequencies, number of rotations, and sinusoidal waves

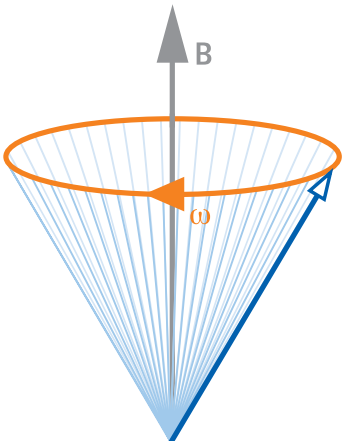
What do we mean by FREQUENCY? It is so-to-speak the "number of rotation" of a periodic movement.

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 revolutions to a time axis we obtain the peaks and crests of a sinusoidal wave. Oscillation with double the frequency is shown as a compressed SINUSOIDAL CURVE.



Precession short and precise

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 precesse relatively slowly at approximately 2,000 Hz (2 kHz).

For MR systems and their high field strengths, spins will precess at a *radio frequency*. This means that the spins precess at several million oscillations per second.

At 1.0 tesla, the Larmor frequency of hydrogen protons is approximately 42 MHz, at 1.5 tesla it reaches 63 MHz. Oscillation frequencies in the megahertz range also include radio waves (AM or FM).

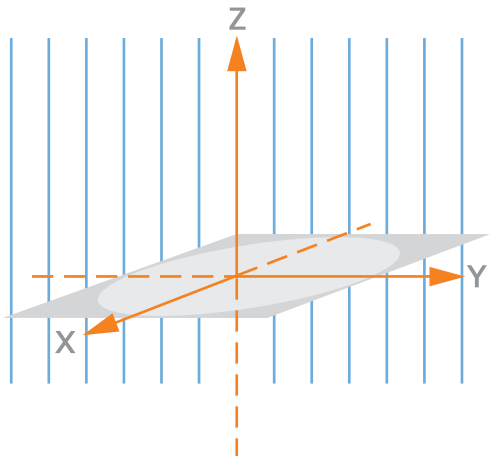
Since the strength of the MR magnetic field is known, the Larmor frequency of the proton spins is known as well. The MR system is tuned to this frequency. The RF coils used build up a "radio connection" with the spins.

The xy-plane

To begin with let's agree on the following conventions:

In the usual xyz coordinate system, we place the Z-AXIS per definition in the direction of the magnetic field.

Additionally, we will call the plane running transverse to the field lines **the** XY-PLANE.

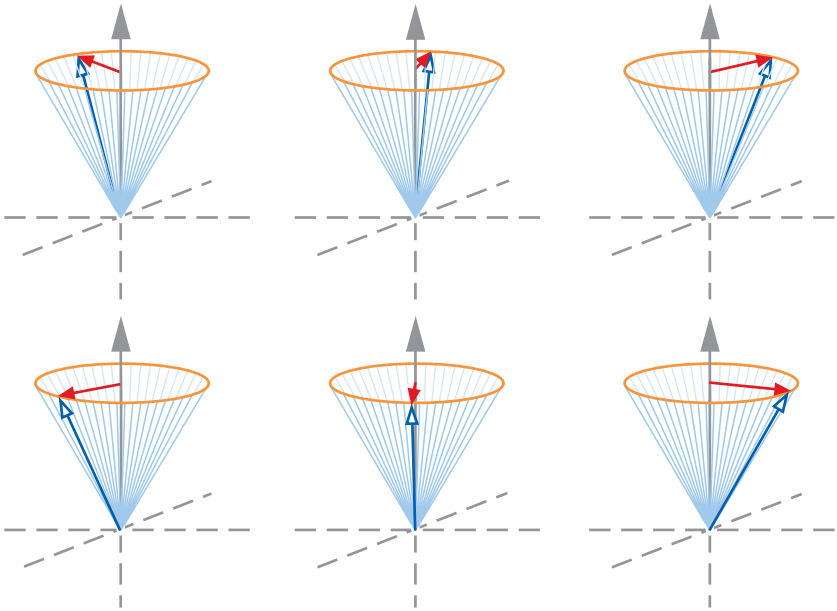


Completely out-of-phase in the basic state

We are going to concentrate on the excess spins of an ensemble and reduce them for clarity's sake into a spinning six pack. All spins precess at the same frequency about the direction of the external magnetic field in completely random orientation.

In other words: while the spins have the same frequency, they are of random → **phase orientation**. And as long as they are of random phase orientation, their components transverse to the magnetic field, that is parallel to the xy-plane, cancel to zero. Constant magnetization **M** will be along the z-axis only.

Spins that are out-of-phase do *not* generate a visible signal.

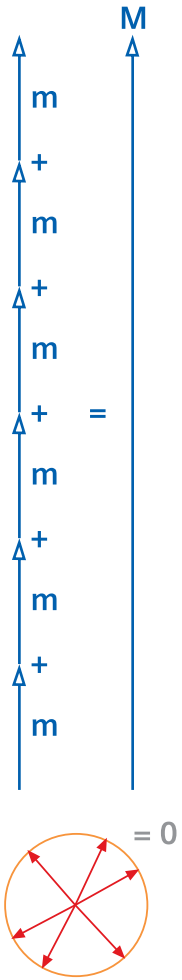


In sum, this is the BASIC STATE of the nuclear spins in the magnetic field:

1. The up and down spins are at equilibrium, the excess spins generate the constant magnetization.
2. The spins precess out-of-phase, their effect is reduced to zero in the xy-plane.

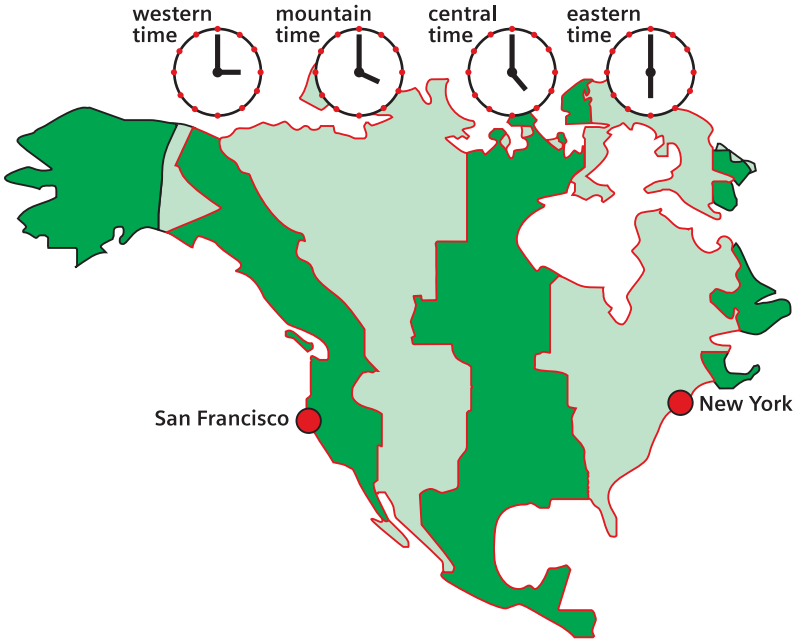
What time is it?

About phases, hands on your watch, and jet lag



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 to another.

If your watch is an hour fast, you could claim that it has a phase shift of 1 hour as compared to local time. You could correct this by setting your watch accordingly, that is, by setting the little 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 known as an irreconcilable phase shift. 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 on we are going to describe how both frequency and phase shifts are used to generate an MR image.



Review

In a magnetic field, the spins precess like tops about the axis of field orientation.

The precession frequency of spin vectors is a function of the magnetic field strength applied. In our case we are speaking about field strengths corresponding to the high-frequency range of radio waves.

Up and down spins are in an energy equilibrium where excess spins create the constant magnetization along the z-axis. The spins precess out-of-phase; their magnetic vectors cancel transverse to the field (xy-plane).

FOR DISCUSSION

Quantum mechanical uncertainties

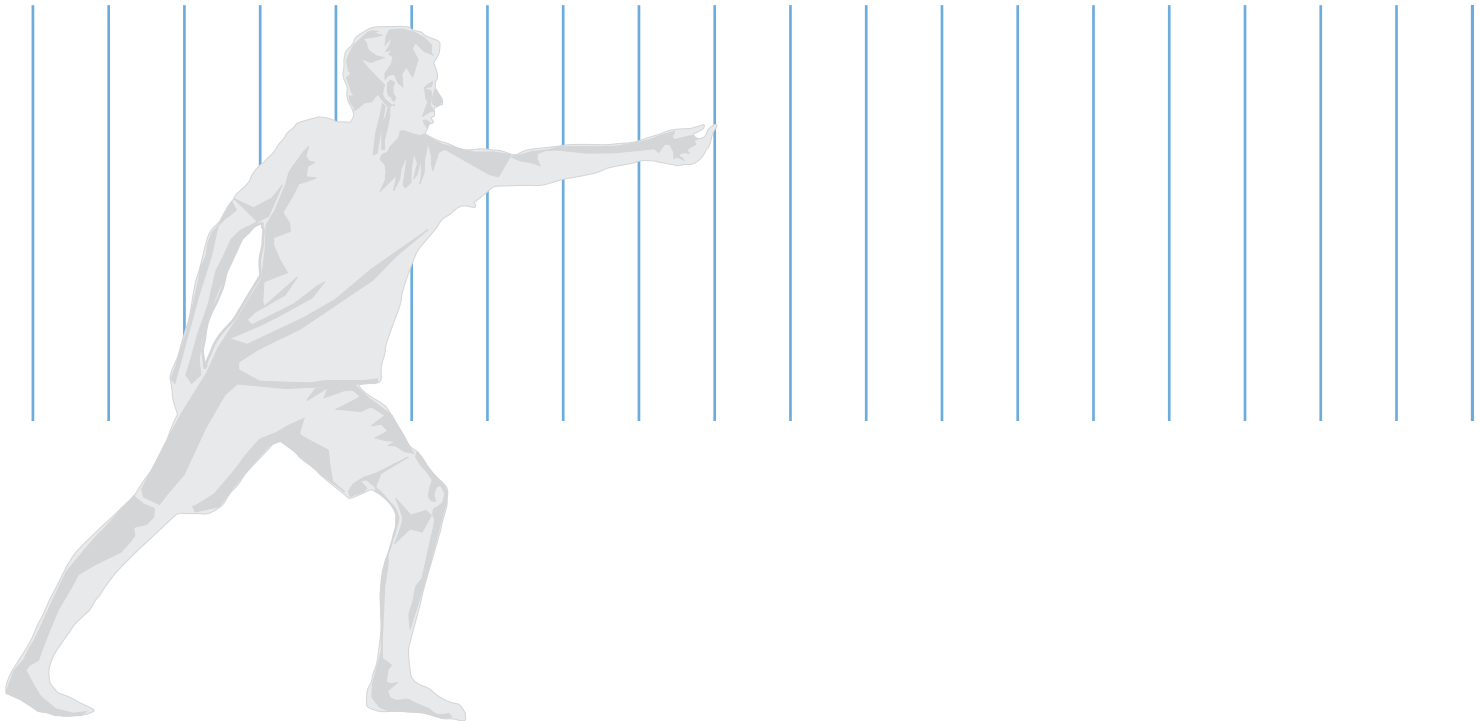
The vector model shows a spin in a superimposed state of up or down that allows for transverse states.

The transverse state of a spin is uncertain when you know its z-component or vice versa.

Due to the uncertain characteristics of the spin states, quantum mechanics works with the *expectation values* of *spin operators*. The expectation value is the average value across a long series of measurements. Fortunately, this value acts like a precessing vector in the magnetic field, allowing us to provide you with the illustration as shown.

Moving spins out of their state of equilibrium

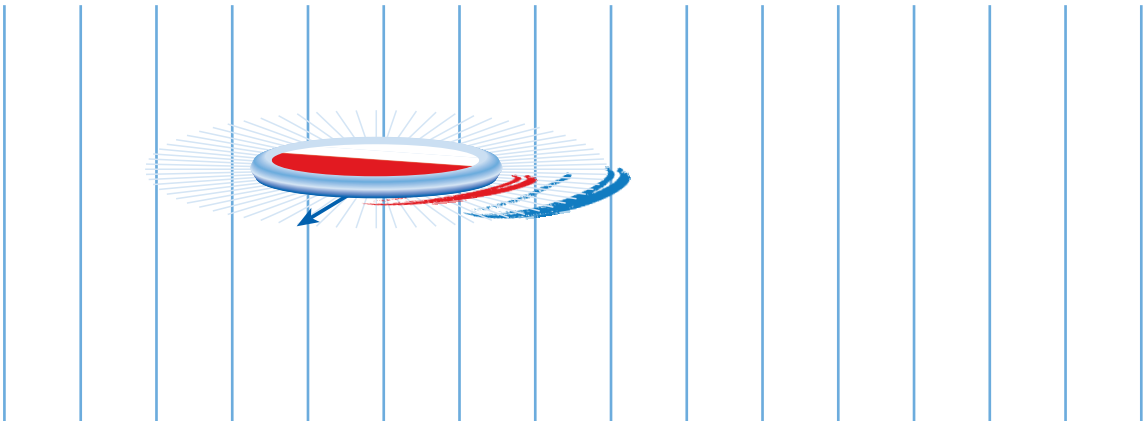
In their basic state, spins precess in the magnetic field in an energy equilibrium, creating a constant magnetization in the body. The nature of magnetic resonance is to deflect the magnetization from its state of equilibrium by interfering with the equilibrium of the spins in a highly targeted manner.



Magnetic frisbees

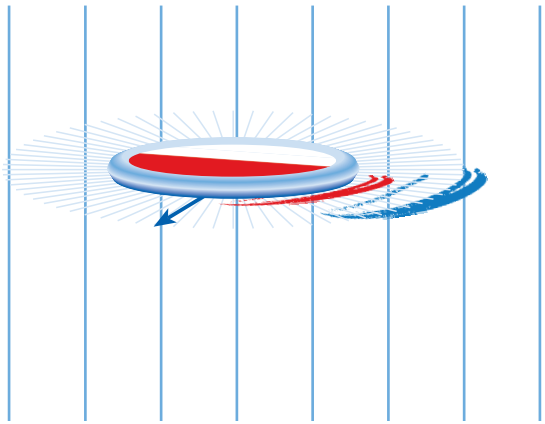
How do you get spins out of their state of equilibrium, how do you change their up and down distribution, their phase locations and their orientation?
One way would be to stimulate them via a magnetic wave. The required short radio frequency wave is also known as the → **RF pulse**.

How do we visualize the RF pulse? One way would be to imagine it as a magnetic frisbee that is suddenly flying through a static magnetic field. What is the frisbee doing? The frisbee acts like a *rotating magnet* that interferes with the homogenous magnetic field.

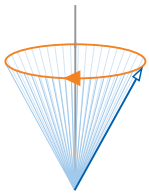


The RF pulse

The state-of-the art RF coils used for MR imaging transmit an RF pulse as a *circularly polarized wave* that contains a rotating magnetic field.



The resonance condition



Why does the RF pulse "interfere" with the spins? Actually, the RF pulse interferes only when the frequency is right.

Important facts are: to interfere with the spins' equilibrium, the RF pulse has to be in → **resonance** with the spins. In other words, the rotating magnet has to rotate at the same speed as the magnetic spin tops.

Physically, the *resonance condition* means:

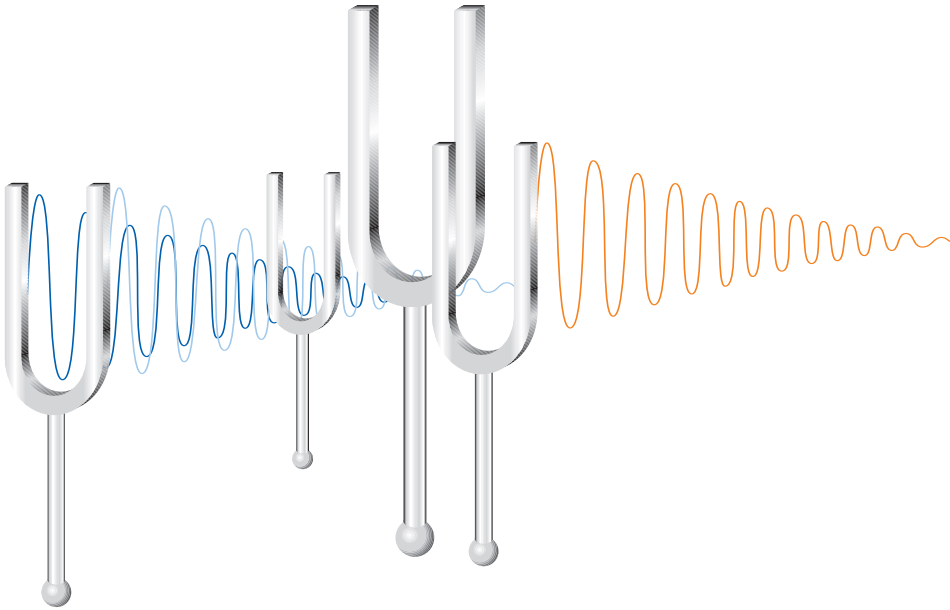
The oscillating frequency of the RF pulse has to match the Larmor frequency of the spins.

Sounds united

Tuning forks in resonance

Resonance stimulation in MR is comparable 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 from the first tuning forks. The two tuning forks are now in RESONANCE.

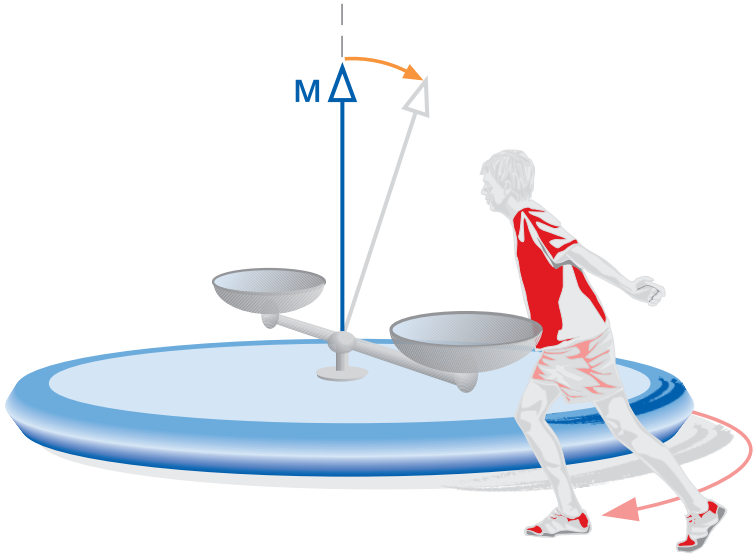


The story of the merry-go-round

What exactly is happening with respect to magnetic resonance? Let's look at yet another analogy:

Imagine yourself as the rotating magnet (that is the RF pulse) that has to be in *resonance* with the rotating spins.

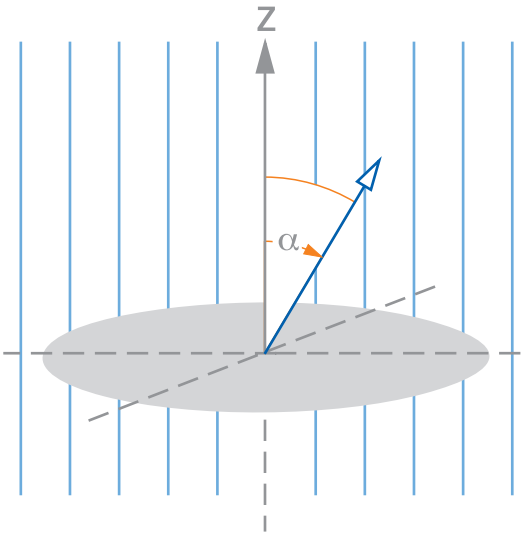
For this purpose, you are running around the spin carousel throwing stones into a rotating "spin scale". Your time is limited. In case you are running either too fast or too slow around the merry-go-round, you are no longer in step. At that point you can only catch up with the scale after one complete 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.



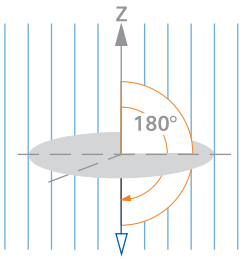
The spin scale loses its equilibrium and the magnetization simply flips over.

Pulses and flip angles

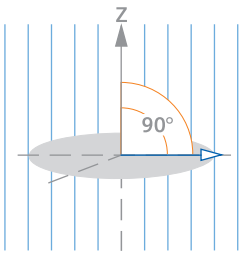
The stronger the energy of the excitating RF pulse, the farther the magnetization will flip or tilt. The final tilt angle is known as the FLIP ANGLE (α).



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**.

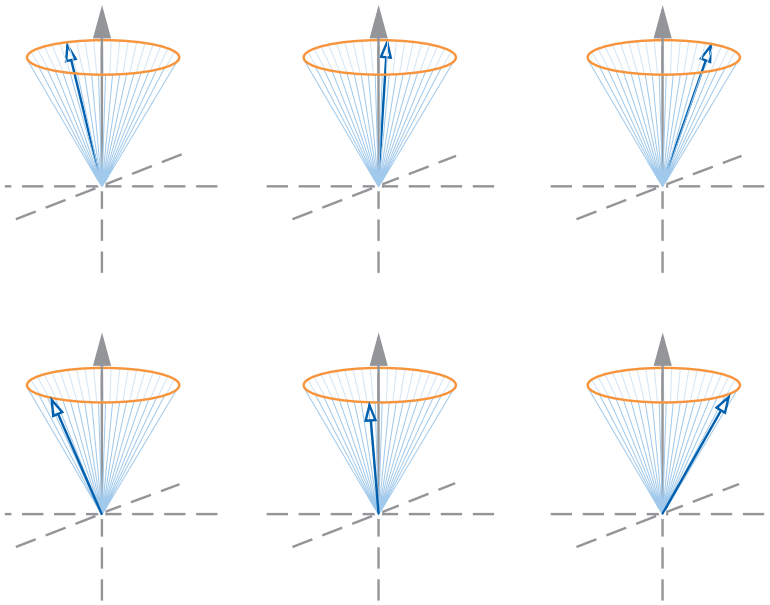


FOR DISCUSSION

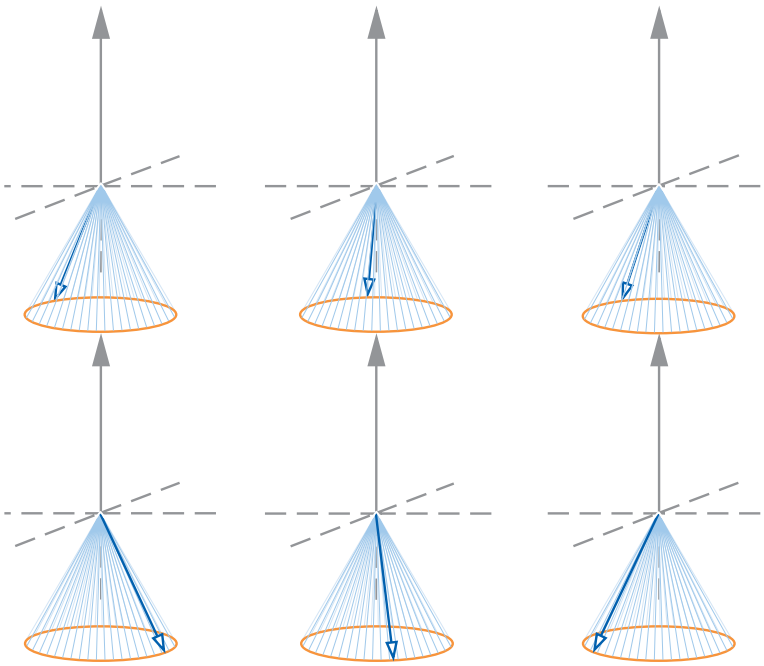
180 degrees—The handstand of the excess spins

Let's look at the flip in magnetization as seen from the viewpoint of the spins.

To explain the effect of the 180 degree pulse, we are going to simplify matters. You are going to be one of the excess spins in our six pack. The RF pulse transfers energy to you, enough to make you do a handstand.



Before the 180 degree pulse



After the 180 degree pulse

It's the same with spins: they flip, that is they jump from the up state into the higher-energy down state (the handstand is unstable as well as in a higher-energy state).

After a 180 degree pulse, all excess spins jumped from the up state into the down state. Net magnetization is now in the opposite direction.

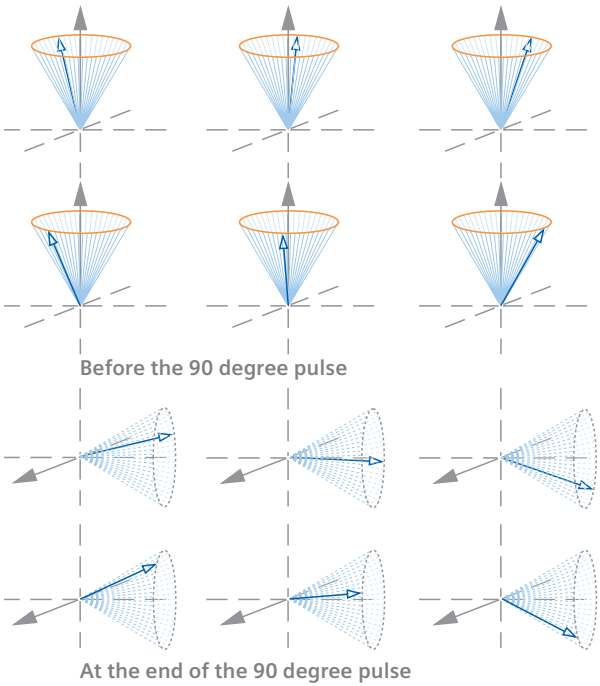
As we will see later, this state is the more unstable one for the spin ensemble as well. It will return again to the energy equilibrium.

90 degrees—To move phases into step

By applying a 90 degree pulse, magnetization flips in the transverse direction, that is, in the xy-plane. And our picture about flipping spins no longer suffices. We need to take a closer look.

As long as the RF pulse is present, two magnetic fields are in effect: the static field, and for a short time, the rotating RF field. By using a special trick, we can let the static field disappear:
all we need to do is climb onto the spin carousel 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 spins, its axis appears *static* to the spins (in our example, the axis points forward). How do the spins react to this magnet vector? They precess about its effective axis.

This is how the original *longitudinal magnetization* in the z-direction is aligned by a 90 degree pulse into the xy-plane. Will the xy phase components of the spins cancel?



Before the 90 degree pulse

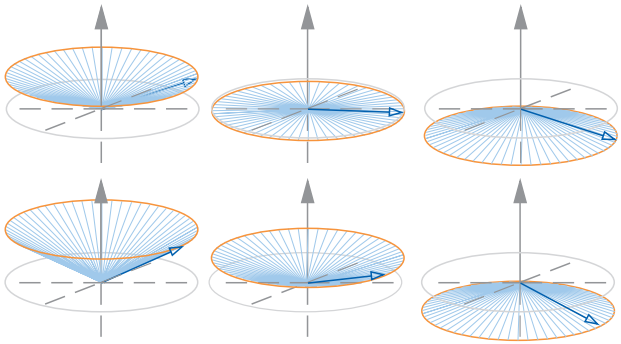
At the end of the 90 degree pulse

Certainly not. Otherwise magnetization would be *zero in all directions* at the end of a 90 degree pulse. But as you can see from our picture, the xy components of the spins don't point randomly in all directions, instead all of them share more or less one direction (to the right in our example).

As we can see, the 90 degree pulse moves the phases of the spin in step. After the uniform spinning of the spin vectors about the axis of the RF pulse, they concentrate in the horizontal direction. This is comparable to the entire six pack lying to the right.

Now to the z components of the individual spins. They statistically cancel across the entire ensemble. The longitudinal magnetization is now zero.

After the pulse, the spins only feel the static magnetic field and continue to rotate about the z-axis. Since they precess with phase coherence (in-phase), they generate a net magnetization in the xy-plane, or a *transverse magnetization* of the same strength as the original longitudinal magnetization. The magnetization is now flipped by 90 degrees.



After the 90 degree pulse

Review

The RF pulse causes the spin ensemble to lose its equilibrium, that is, the RF pulse meets the following requirements: the oscillation frequency of the RF pulse matches the Larmor frequency of the spins.

A 90 degree pulse will flip magnetization into the xy-plane. A 180 degree pulse will flip magnetization into the opposite direction of the z-axis.

How do we obtain an MR signal?

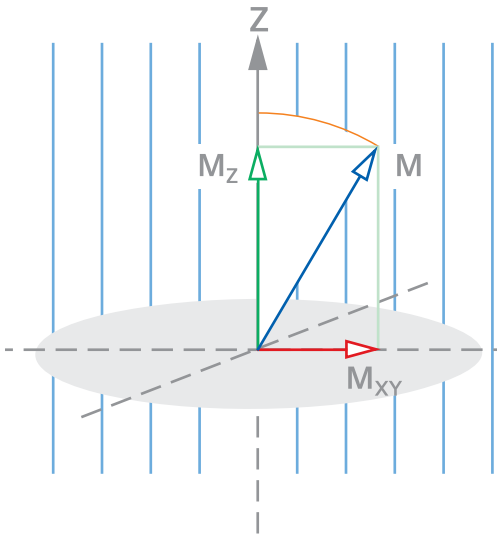
The RF pulse interferes and the magnetization flips, creating a component in the xy-plane. How does the flipped magnetization generate a signal?

Decomposing the magnetization

Just like a vector, magnetization can be decomposed into two components located vertically 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 spin vectors that rotate in-phase in the xy-plane, matching Larmor frequency. Ergo, transverse magnetization also rotates with the Larmor frequency.



The MR signal generated by transverse magnetization

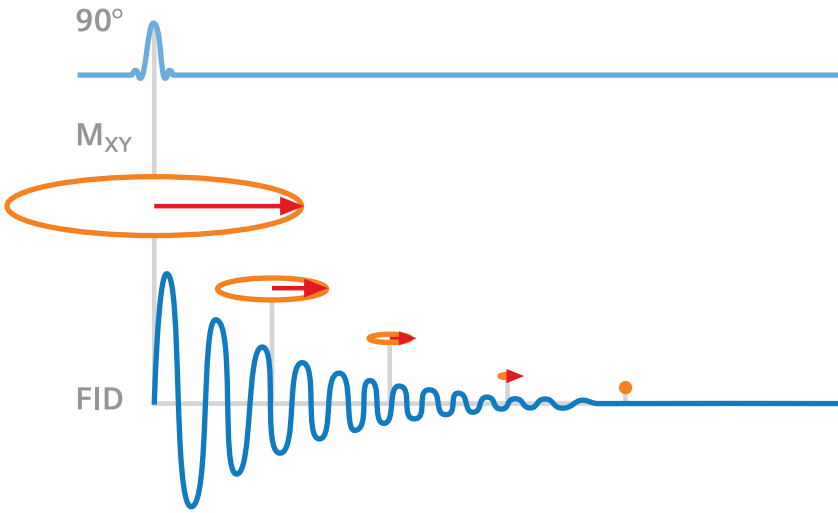
The transverse magnetization acts like a rotating magnet. You can move a coil into the rotating magnet and induce an → **electric voltage** in it.

The course of the voltage over time is the MR SIGNAL. The stronger the transverse magnetization, the stronger the MR signal. Note that it 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 the FREE INDUCTION DECAY or **FID**.



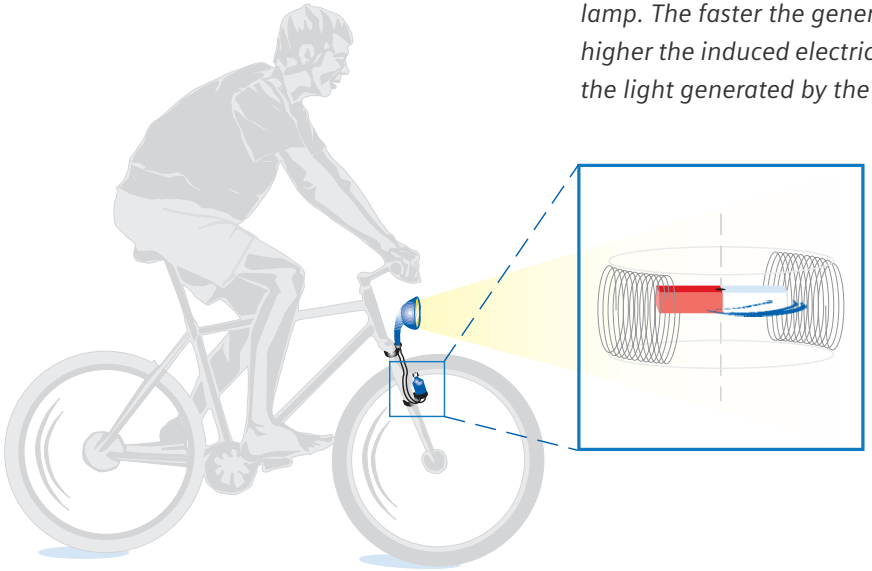
The reasons for signal decay are described in the next chapter.

The voltage is increasing

What you should know about electromagnetic 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 generator contains a rotating magnet driven by the wheel of the bicycle, continuously changing the direction of its magnetic field. These changes in magnetic field induce a flow of current in the coil which may be used to light up the bicycle lamp. The faster the generator magnet rotates, the higher the induced electric voltage and the brighter the light generated by the bicycle lamp.



Summary

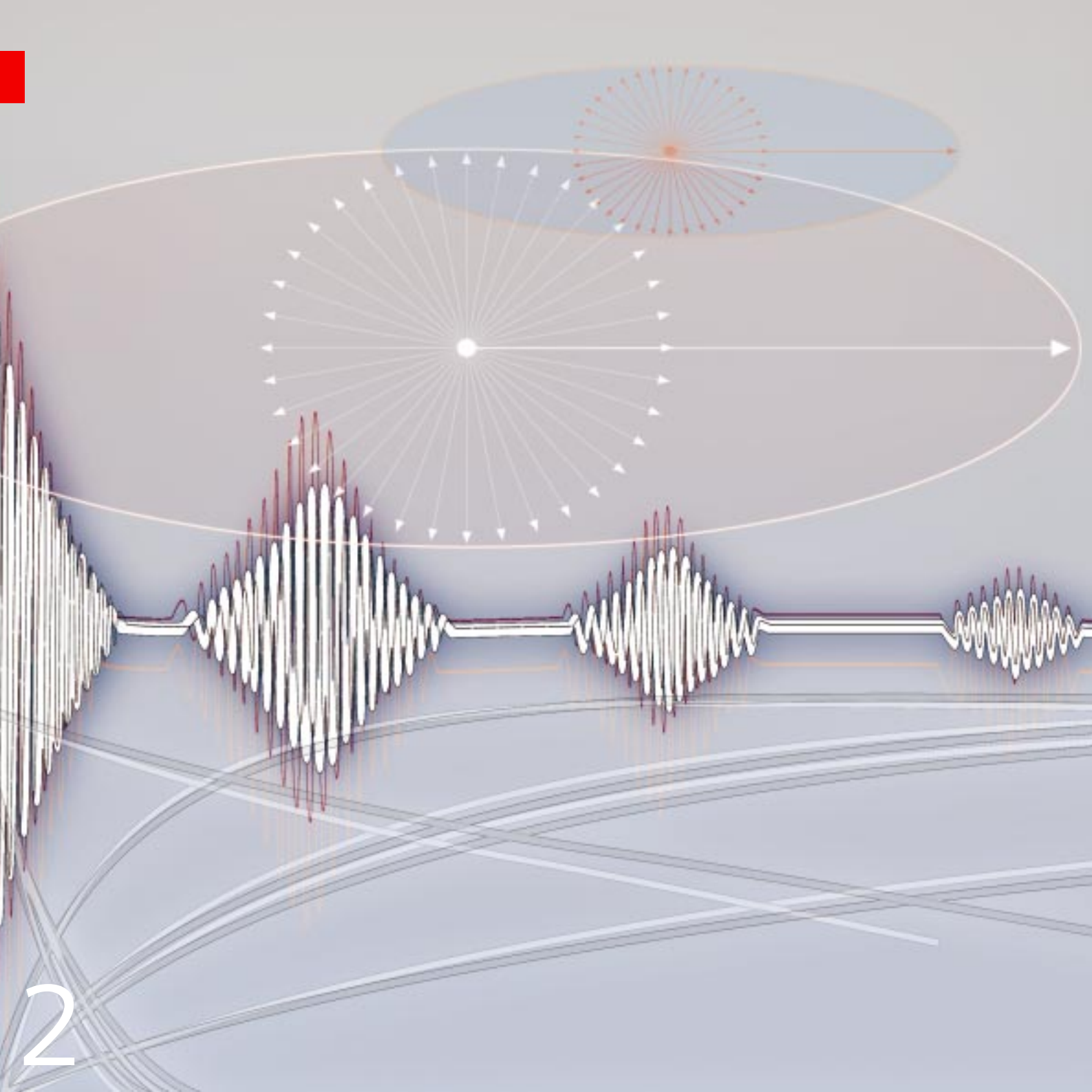
As a first step, we have discussed the magnetization of non-magnetic probes in a static magnetic field. Magnetization in the direction of the external field is build up.

The energy equilibrium is *dynamic*: the individual nuclear spins spontaneously change their energy state. The total number of excess spins, however, remains the same and thus constant magnetization is maintained.

The RF pulse applied causes the spin ensemble to lose its original equilibrium.

After the end of a 90 degree pulse, magnetization is flipped into the xy-plane and rotates at the Larmor frequency.

The rotating transverse magnetization generates the MR signal and decays quickly (FID).



About spin relaxation and echoes

The longitudinal magnetization is zero after a 90 degree pulse and rotates as transverse magnetization in the xy-plane.

Will it continue in this state? No.

The transverse magnetization is lost rather quickly and as a result, the MR signal drops.

What does occur is that after the 90 degree pulse, the longitudinal magnetization recovers fully as if nothing had happened.

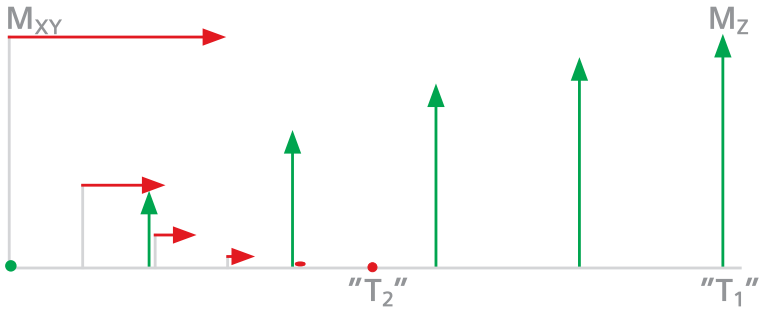
This process is known as RELAXATION.

Understanding relaxation

After each interference caused by an RF pulse, the spins return to their basic state again, that is, they recover. RELAXATION, as we will discover, can be described by two independent processes by viewing longitudinal and transverse magnetization as separate entities.

Longitudinal and transverse

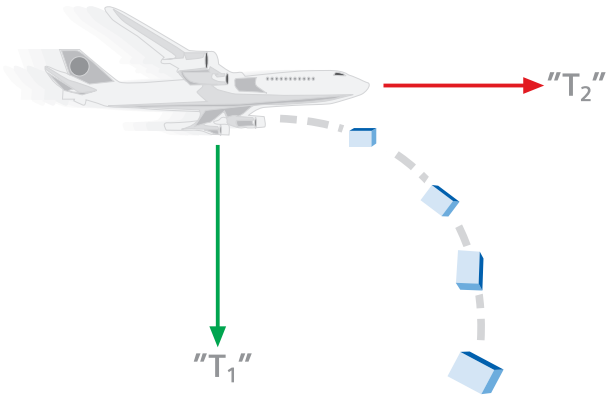
It's plausible to think the following when transverse magnetization decays and longitudinal magnetization recovers: magnetization, left alone, will return to the z-direction.



However, this is not true.

Transverse magnetization M_{xy} decays more rapidly than the time required by longitudinal magnetization M_z to recover. Both processes run → **exponentially**.

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



A falling box

We can compare this 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.

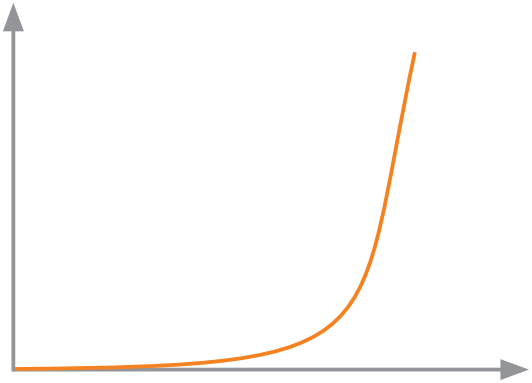
If the box is thrown from an airplane, two simultaneous "forces" are at work: 1. gravity, 2. the kinematic energy in the direction of flight.

The actual movement of the box is the *superposition* of two motions performed independent of each other. While the box is traveling toward the earth, coming closer and closer, it barely continues in the direction of flight.

Up the mountain, down the mountain

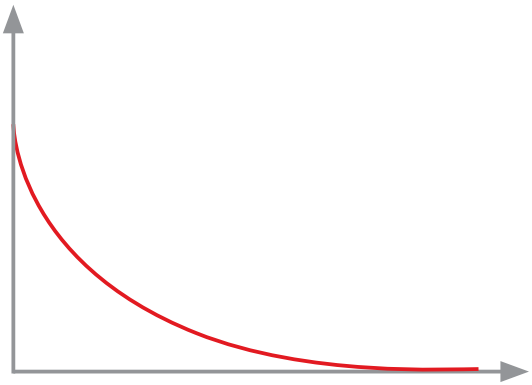
Interest compounded and exponential growth

Many natural as well as well as social processes can be expressed mathematically in a rather easy way: they are EXPONENTIAL. The increase in bacteria, 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's assume, you have stocks or fixed rate funds in the value of $ 10,000 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 for exponential decay. Starting out by assuming a cash balance of $ 100,000 and a rate of inflation of 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.

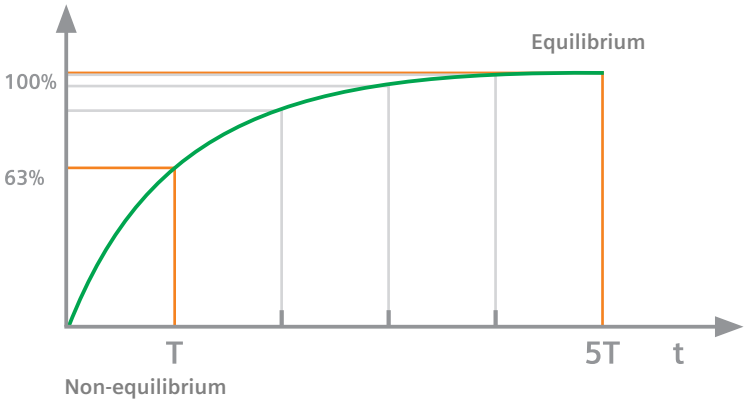


What do we mean by relaxation?

RELAXATION is a dynamic process:
a system returns from its state of non-equilibrium to its state of equilibrium.

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

You can compare this to a rubber band. The tension of the rubber band determines how quickly it will return to its normal state.

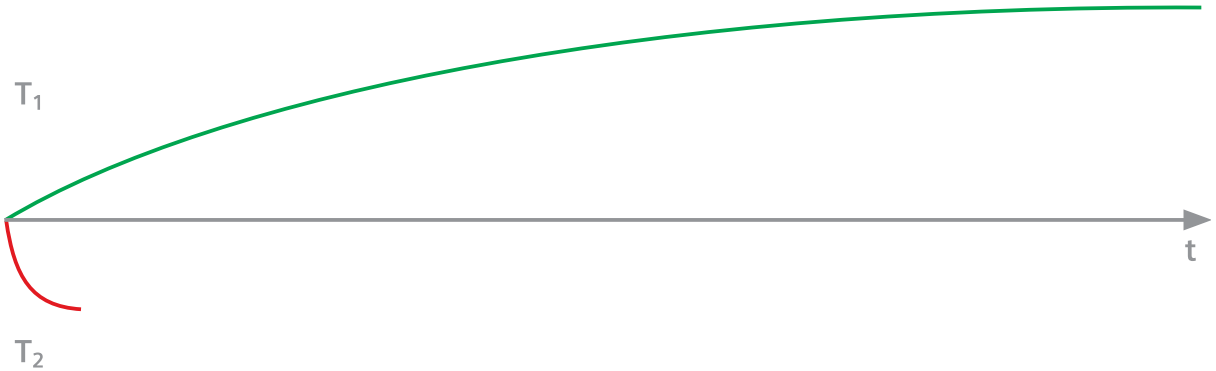


An exponential process is expressed by its TIME CONSTANT T.

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 and the quantity has reached its state of equilibrium.

It's faster getting down the mountain than getting up

Let's summarize: while longitudinal magnetization is recovering, transverse magnetization is decaying. And as mentioned before, the decay of the transverse magnetization is *much quicker* than regrowth 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—up the mountain—T_1
Transverse—down the mountain—T_2

Recovery of the longitudinal magnetization (T_1)

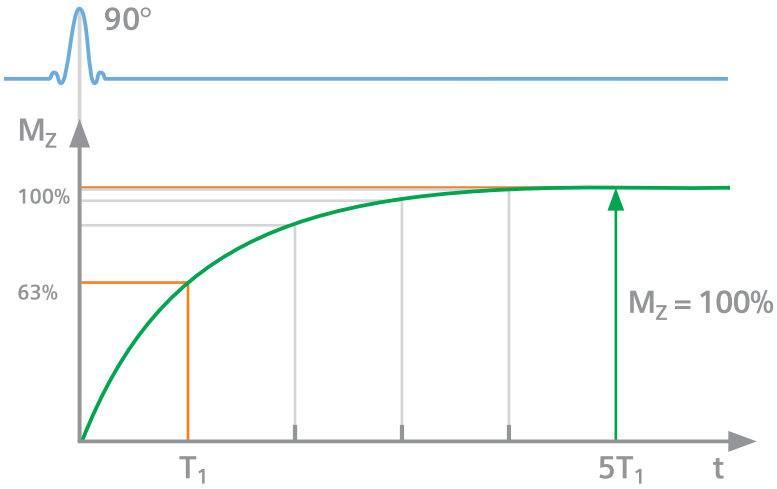
After a certain time period, the longitudinal magnetization has fully recovered from the RF pulse. The spin ensemble in the static magnetic field is moving toward its energy equilibrium.

Return to equilibrium

The recovery of the longitudinal magnetization is an exponential process, known as LONGITUDINAL RELAXATION. The time constant is known as T_1.

After T_1, the longitudinal magnetization M_z has recovered to approximately 63 % of its final value. After 5 T_1 times, recovery is complete.

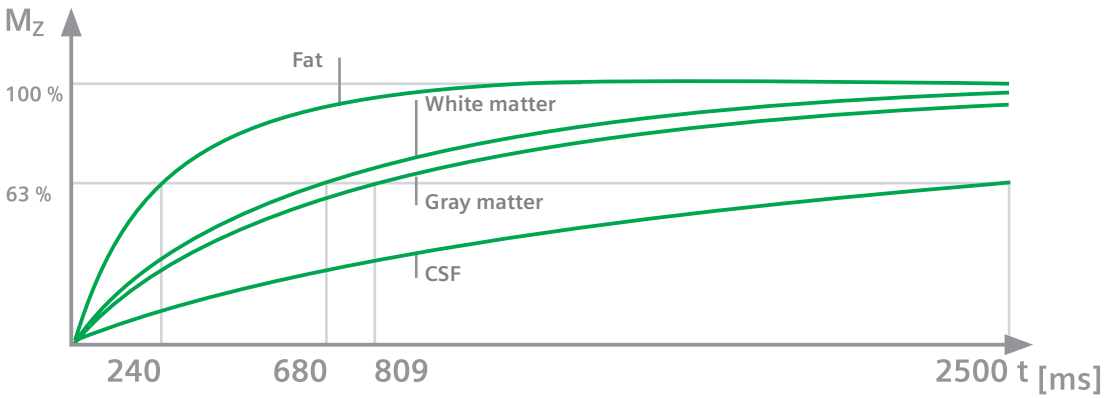
Is T_1 the same everywhere? Everywhere in the body, for all tissue types? Fortunately, it is not. The T_1 constant depends on the tissue, that is, it's tissue-specific.



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.

Why is that?
The RF energy of the stimulated spins is lost again through interaction with the → **lattice**.



As the table shows, the T_1 constant *depends on the field* as well.

T_1 constants (in ms)

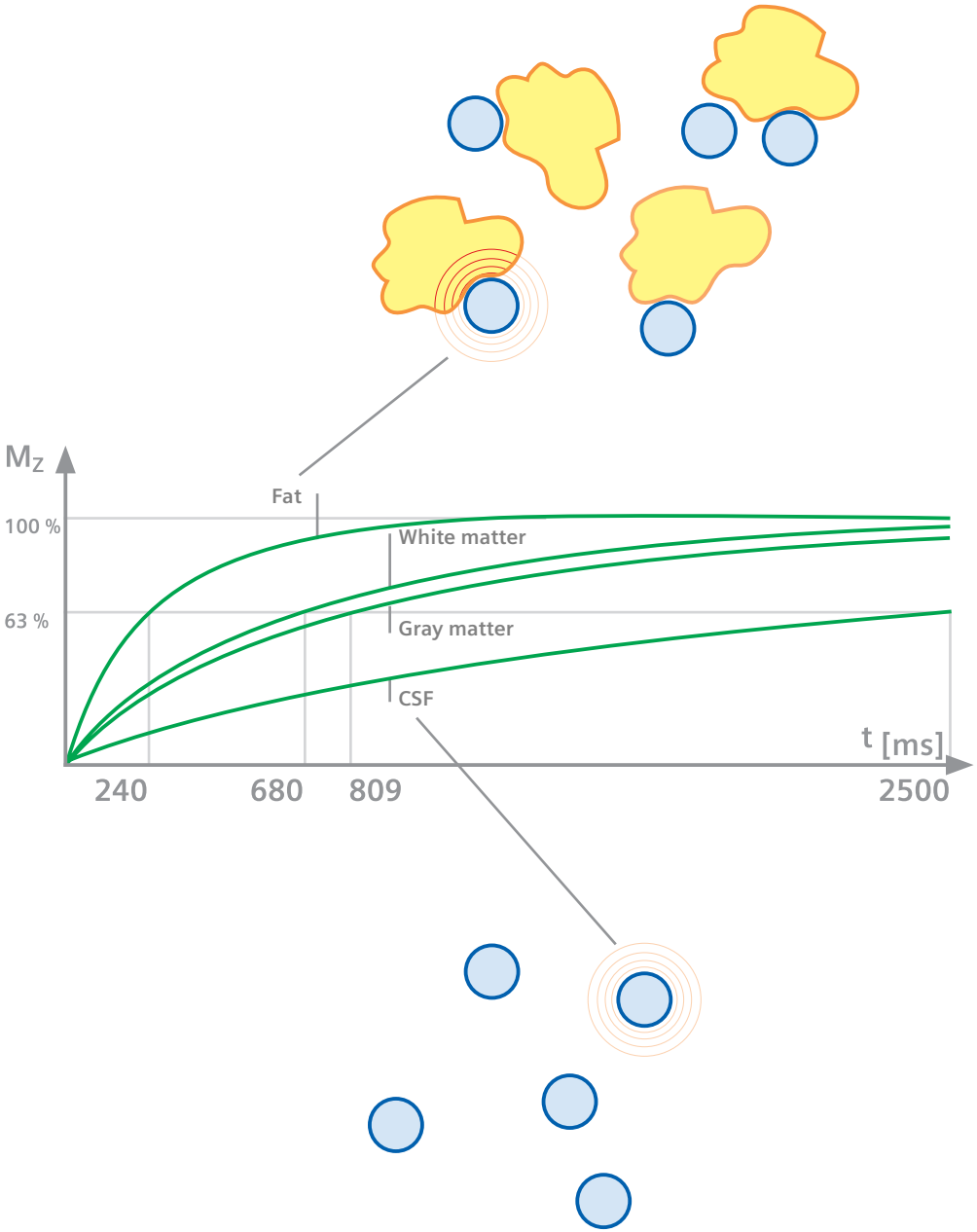
	0.2 Tesla	1.0 Tesla	1.5 Tesla
Fat		240	
Muscle	370	730	863
White matter	388	680	783
Gray matter	492	809	917
CSF	1,400	2,500	3,000

Rule of thumb:
Fat has a short T_1,
water has a long T_1.

FOR DISCUSSION

Spin-lattice relaxation

The protons change their spin status at resonance. What makes them return to equilibrium at the end of the RF pulse? Actually, the protons continuously "feel" local fields and their fluctuations caused by molecular motion (magnetic noise). These minute magnetic field fluctuations are superimposed on the external field. The strongest effect is created by magnetic field fluctuations that match the *Larmor frequency* of the protons and oscillate transverse to the main field. They behave like small RF pulses and cause the spins to "flip".



The environment of the protons frequently consists of larger molecules (lipids) and macro-molecules (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 state. This explains the relatively *short* T_1 relaxation of fatty tissue, for example.

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

How are these field fluctuations generated? They are generated by magnetic dipole fields of unpaired electrons as well as other nuclei.

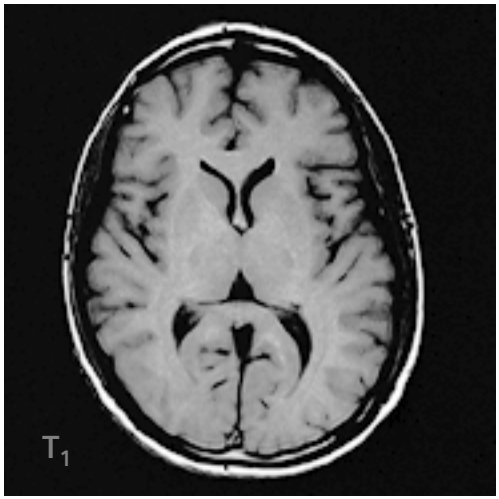
The environment of a proton is called "lattice". Although originally reserved for lattice structures in solids, the term applies to liquids as well. Since the spin ensemble emits energy to the lattice during longitudinal relaxation, the T_1 process is frequently called the SPIN-LATTICE RELAXATION. This process occurs after interference from an RF pulse and as early as during the building-up of the longitudinal magnetization, after the patient has been moved into the magnetic field.

So far we have talked about the T_1 constant which is a function of the size of the tissue molecule, its mobility and its type of environment. The T_1 constant indicates how quickly a spin ensemble in a certain type of tissue will be able to emit its excess energy to the lattice.

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 difference in relaxation is visualized as contrast in the MR image.



With T_1 contrast CSF appears as dark in the MR image

Review

After an interference the spin ensemble returns to its energy equilibrium. The longitudinal magnetization fully recovers within seconds. This process is known as *longitudinal relaxation*.

The longitudinal relaxation follows an exponential course of growth characterized by the time constant T_1. T_1 is a measure for the restoration of the longitudinal magnetization.

The T_1 constant is tissue-dependent. This characteristic is employed for the MR image contrast.

The source for T_1 relaxation are local magnetic field fluctuations created through molecular motion. Magnetic field fluctuations in the range of the Larmor frequency show the strongest effect and cause the protons to change their spin state.

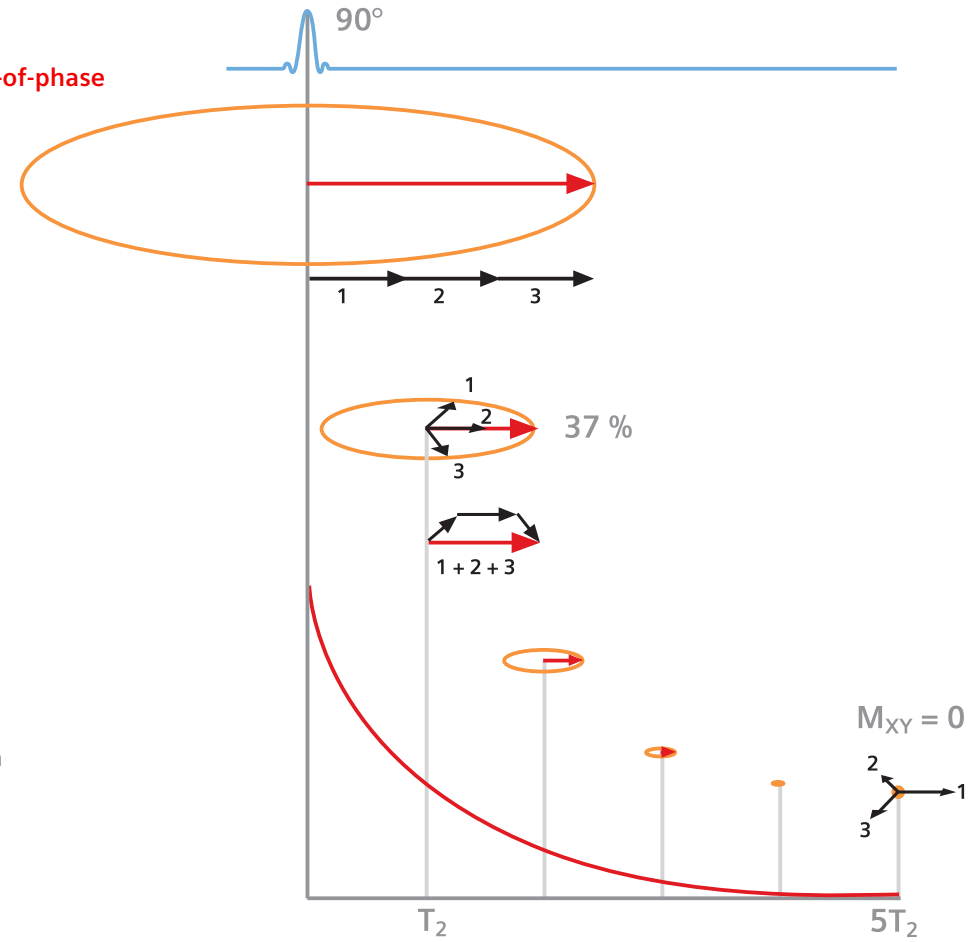
Transverse magnetization is decaying (T_2)

After a 90 degree pulse, the subsequent rotating transverse magnetization generates the MR signal. This signal, the FID (free induction decay), decays quickly. In other words, the transverse magnetization is lost again and the spins are once more out-of-phase.

The spins get out-of-phase

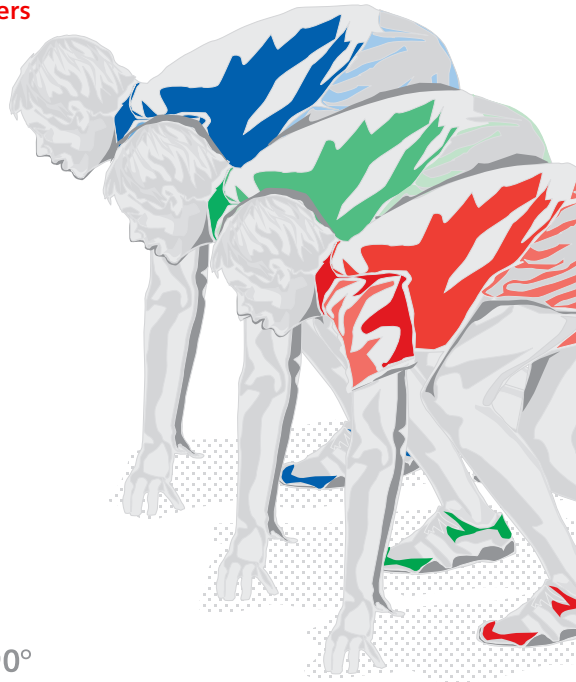
Directly after the RF pulse, the spins are what is called *phase-coherent*. They act like one large magnet that rotates in the xy-plane.

However, the rotating spins lose their coherence again due to unavoidable interactions. The spins get out-of-phase again and the transverse magnetization begins to decay.



The example of the runners

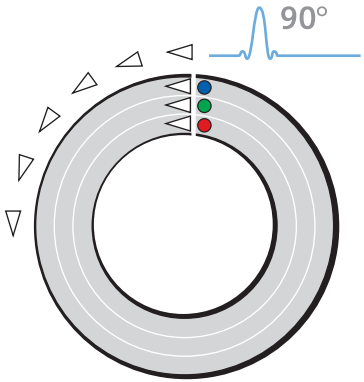
We can compare this to a group of runners. At the beginning of the race, they are all lined up at the starting line.



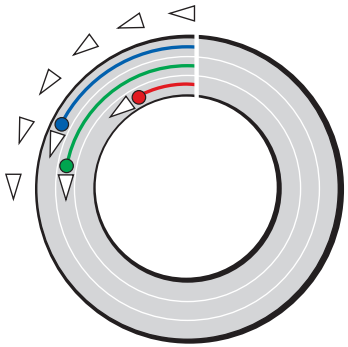
The following is crucial for understanding MR imaging: the spins DEPHASE, that is, the rotating transverse magnetization fans out into its individual spins and starts to decay. The MR signal dies out.

This is what we call TRANSVERSE RELAXATION. Its time constant is known as T_2. As we will see later on, this is the ideal time only. In reality, the FID decays at a more rapid rate.

After 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.

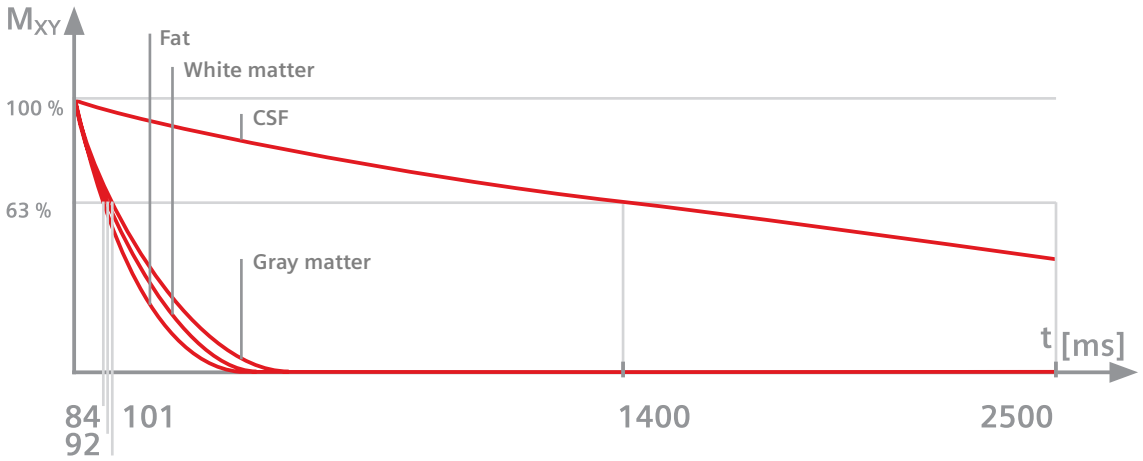


During the race, the group of runners fan out more and more since they are running at different speeds. You, as the spectator, will see that the order among the runners at the starting line—let's call that order coherence—gets lost during the race.



The T_2 constant under the magnifying glass

T_2 is also tissue-specific.



T_2 constants are largely field-*independent*.

In sum, the same applies as for T_1.

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.

Where does transverse relaxation differ?

The relaxation processes that determine the increase in longitudinal magnetization also determine the decay of the transverse magnetization (comparable to the falling box which is always subject to the earth's gravity). Since transverse magnetization decays more quickly than longitudinal magnetization regrows, the decay must be due to an additional mechanism (the box is dropped at the speed of the plane).

The additional processes are in the main the → **spin-spin interactions** inside the ensemble.

FOR DISCUSSION

Spin-spin relaxation

Although the interaction between the spins is not the only source for transverse relaxation, the term SPIN-SPIN RELAXATION has made its entrance and is here to stay.

As we have shown, fluctuating magnetic fields close to the Larmor frequency are responsible for protons changing their spin states. This is the cause for longitudinal relaxation. And it has its transverse effect: whenever the spin state changes, the phase gets lost as well. Flipping spins lose their phase coherence, our spinning tops begin to dephase. This means that the dynamic processes of the longitudinal relaxation cause the transverse relaxation as well.

In addition the change in spin state also changes the local field by a small amount. The z component of the spin is now pointing into the opposing direction. Adjacent protons feel a local change in the magnetic field in the z-direction. This change amounts to approximately 1 milli tesla.

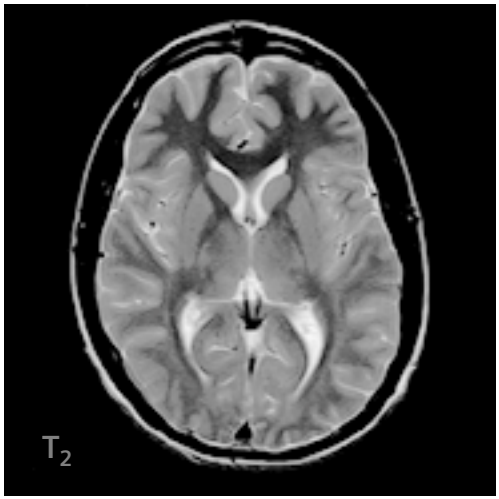
What does this mean for the spins? When the static magnetic field shows local differences, the precessional frequency in this area differ as well. For this reason, the variance of precessional frequencies of stimulated spins is approximately 40 kHz around the normal Larmor frequency.

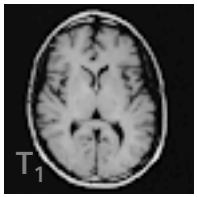
Because of these slightly different precession frequencies, the rotating spin magnets are no longer in step. Just like runners moving at different speeds. Their common effect disappears before recovery of the longitudinal magnetization.

A voxel may contain different tissue types. The transverse relaxation is then the result of complex interactions and it is difficult to describe it as a simple exponential curve.

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 as bright in the MR image—opposite to T_1 contrast

Review

Immediately after stimulation with an RF pulse, the nuclear spins return from their excited state into their basic state:

1. The energy equilibrium between up and down spins is restored, the excess spins generate the longitudinal magnetization.

2. The spins rotate out-of-phase, and it is not possible to observe the transverse magnetization.

The transverse relaxation follows an exponential decay curve characterized by the time constant T_2 which is a measure of the nuclear spin dephasing.

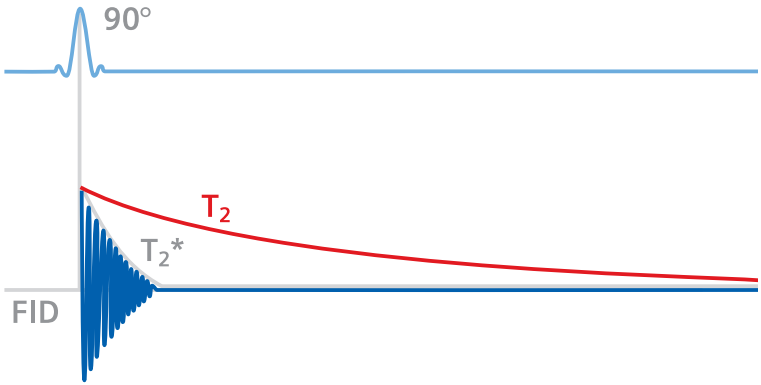
The T_2 constant is also tissue-specific and contributes to the image contrast.

Spin echo (T_2*)

The MR signal has decayed and so has apparently. the transverse magnetization. This is when the magical moment comes into play: we get the MR signal back and this wonderful trick allows us to generate a spin echo.

The true decay of the FID

Actually, we could expect that the MR signal (FID) decays with the constant T_2. However, the FID decays much more quickly, that is, with a shorter *effective* time constant T_2*.

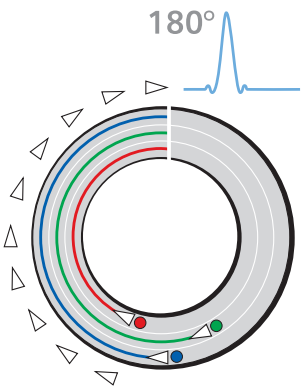


Why is that?

The static magnetic field as felt by the spins is not the same everywhere. In other words, it is INHOMOGENEOUS. As compared to the 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 local field variations caused by the patient's body as well as technical inhomogeneities of the magnet.

It is these static magnetic field differences that add to the fanning of the spins: they dephase more quickly than the T_2 relaxation.

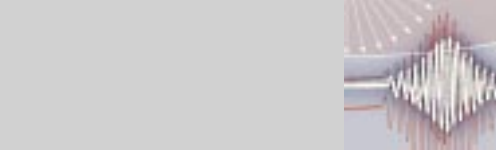


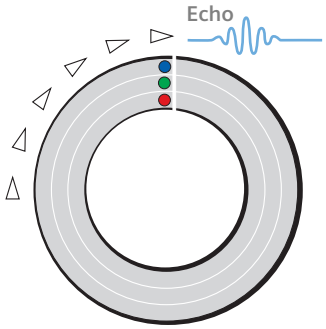
The about face trick

Why are we getting involved with the T_2 constant? Especially since the phase coherence of the spins seems to be irrevocably destroyed in the T_2* time.

Well, we are not correct in our assumptions

Do you remember the runners fanning out? We can bring them back into step: after a certain time period we will ask all runners to turn "about face". By that we mean that they should turn around by exactly 180 degrees and then continue running.





The first will be the last...

The fastest runners are now the slowest or last runners. However, as a prerequisite they have to be running continuously at the same speed. If this is the case, they will arrive within the same time span as he slower runners a the start line. You can compare this to a film that is running in reverse.

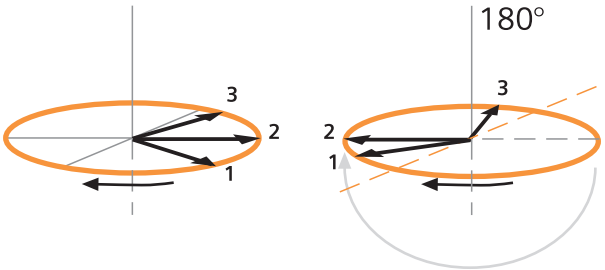
As a spectator, you may have thought that the order at the start line would have been completely destroyed during the race. But now you can see that order has been restored with our “about face” trick. We are experiencing an “echo” of the start.

Spins flipped like an omelet

Since static magnetic field differences remain constant with respect to space and time, we can cancel their effect through an about face trick as well.

We don't proceed in the same way as for the runners. Otherwise, we would have to reverse the polarity of the magnetic field (all spins would rotate in the opposing direction).

Instead we give the "about face" command via a 180 degree pulse. By means of the 180 degree pulse, the spins are flipped just like an omelet: the order of the spins is reversed, the direction of rotation remains.

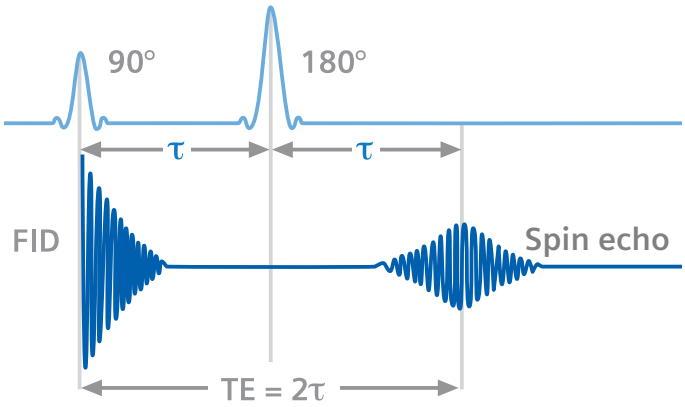


The faster spins (1) catch up with the slower (3), precessing spins...

Here comes the echo

This is the effect of the 180 degree pulse: the out-of-phase spins are going back into phase, and a new MR signal is generated—the SPIN ECHO.

The 180 degree pulse is switched behind the 90 degree pulse after the run time τ. The spin echo signal is initially increasing 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.

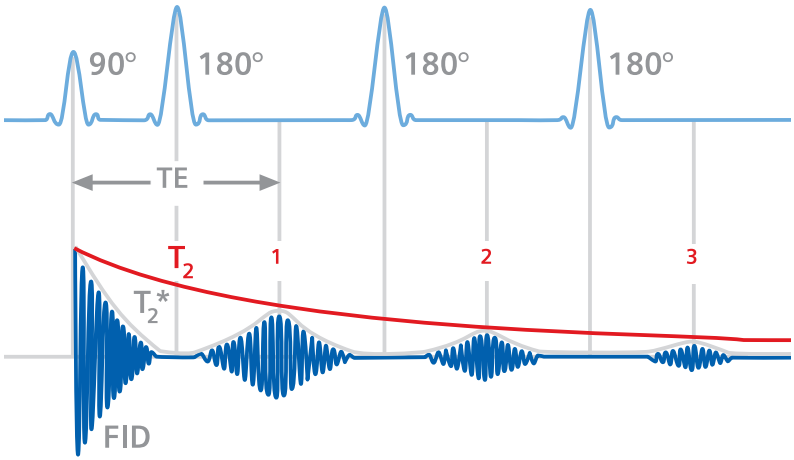


Echoes following one another

When several 180 degree pulses are following each other in sequence, several spin echoes are generated by a MULTI-ECHO SEQUENCE. The amplitude of the echoes is smaller than that for the FID. The larger the echo time, the smaller the echo. We can repeat this until transverse magnetization is irrevocably lost through T_2 relaxation.

Please note: The spin-echo signal decreases with T_2*, its strength (amplitude, maximum), however, decreases with T_2.

Rule of thumb:
$T_2^* < T_2 < T_1$



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

Review

The FID decays with the very short time constant T_2^*. The cause for this fast decay are static magnetic field differences that remain constant over time within a specific location. They allow for a quick dephasing of the spins.

We can get the MR signal with a 180 degree pulse. The MR signal is known as the spin echo.

By applying several 180 degree pulse in sequence, we generate multiple echoes. This is possible as long as T_2 relaxation continues.

Rule of thumb:
$T_2^* < T_2 < T_1$

It is difficult to measure the strength of the FID. For this reason echoes are the preferred signals for imaging.

The gradient echo

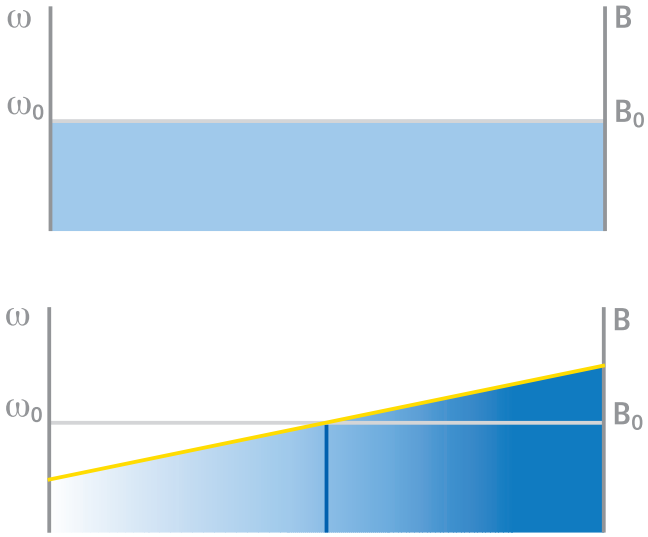
**A FID echo may be generated in a number of ways.
MR imaging uses two basic methods. We have already talked about the spin echo and would now like to go over to its "brother": the gradient echo.**

Changing the magnetic field

Let's forget about the flipped 180 degree pulse for the time being. As a result we do not have a spin echo. How do we then manage to get an MR signal?

We are going to change the magnetic field directly after the RF pulse. The change involves that the field gets smaller in one direction and larger in the opposing one. This change is known as a → **gradient**.

The original field strength B_0 is present in one location only, before and after this location, the field strength is lower or higher. The precession frequency of the spins, as you may know, is directly proportional to the field strength. The spins are now rotating at different speeds along the field change.

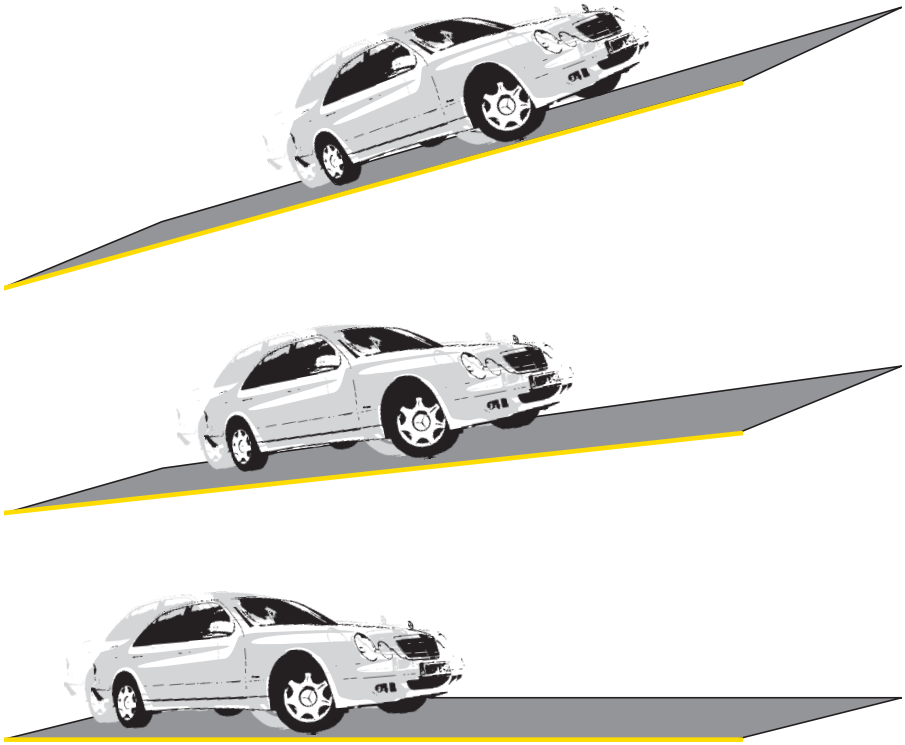


Increasing fields

What is a gradient?

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.

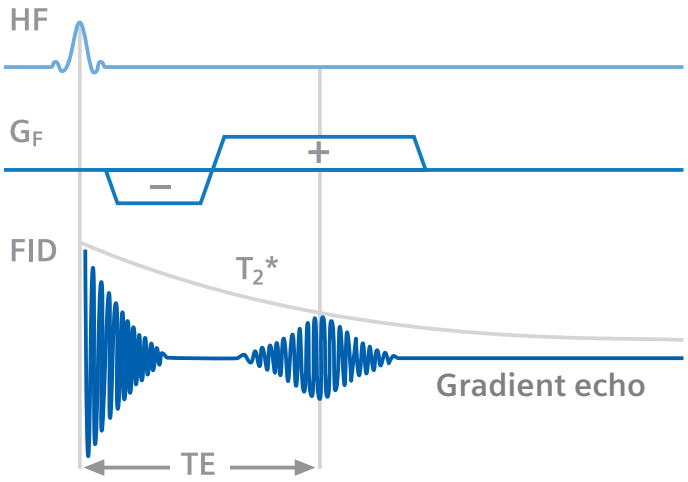
Transferred to MR technology this means: a MAGNETIC FIELD GRADIENT is a change of the magnetic field in a certain direction, that is, a linear increase or decrease.



A simply different echo

A gradient pulse (–) directly after the RF pulse artificially dephases the spin frequencies. Since they are now rotating at different speeds, they lose their phase more quickly, that is, they are being DEPHASED. The FID is destroyed considerably faster than it would under normal conditions.

By reversing the polarity of the gradient (+), the 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 has to be considerable shorter for a gradient echo sequence than for a spin echo technique. Why is that?

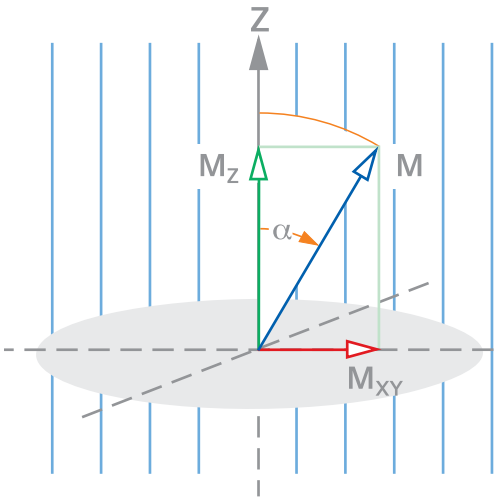
The 180 degree pulse is omitted in gradient echo technology. This means that we do *not* cancel the static T_2* dephasing mechanism as we do in spin echo technology. Instead we use gradient pulses to quickly destroy the FID and build it back up again, all within the T_2* decay.

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.

Reducing the flip angle

To generate gradient echoes, the flip angle for the stimulating RF pulse is usually less than 90 degrees. The advantages of this approach include stronger signals as well as reduced measurement times.

We'll provide more detailed information in the section on gradient echo sequences.

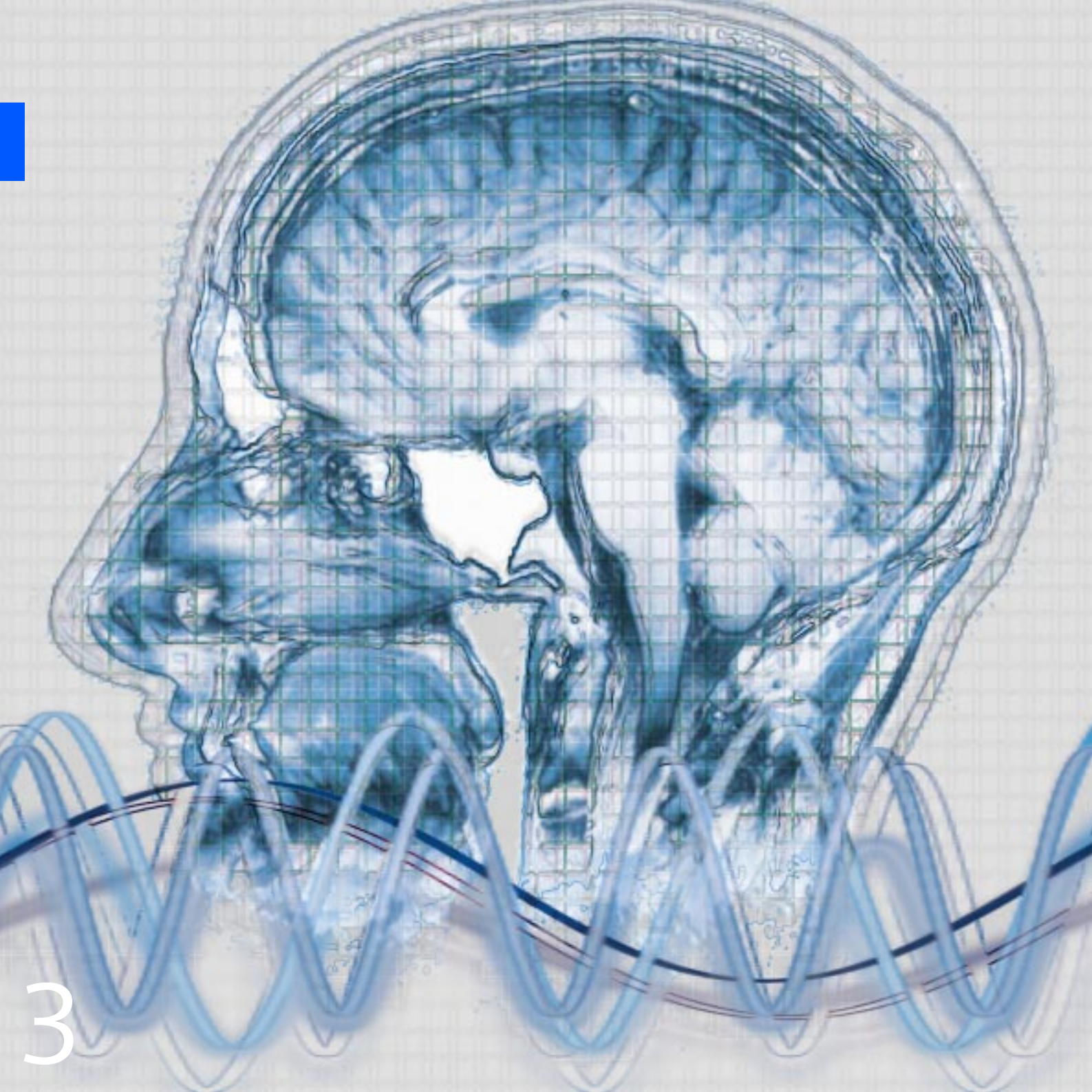


Review

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 faster than the spin-echo technique.



From the signal to the image

During a simple MR experiment we receive an MR signal, be it as a FID, as a spin echo or as a gradient echo. This signal is the sum of all nuclear magnetic resonances in the entire body. However, since we do not have a spatial allocation, we cannot distinguish between the different tissue structures.

But we are nevertheless interested in the following: how do we generate an image from an MR signal that shows spatial structures as different gray values.

The slices that give us the images

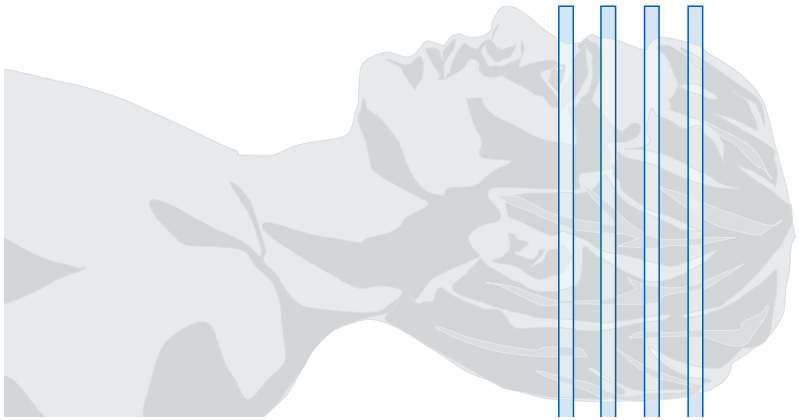
The basis for an MR image is the spatial allocation of individual MR signals that reflect the respective anatomical structure. The common method is to spatially vary the magnetic field. The nuclear spins will then show different precessional frequencies at different positions. Magnetic resonance is now spatially differentiated.

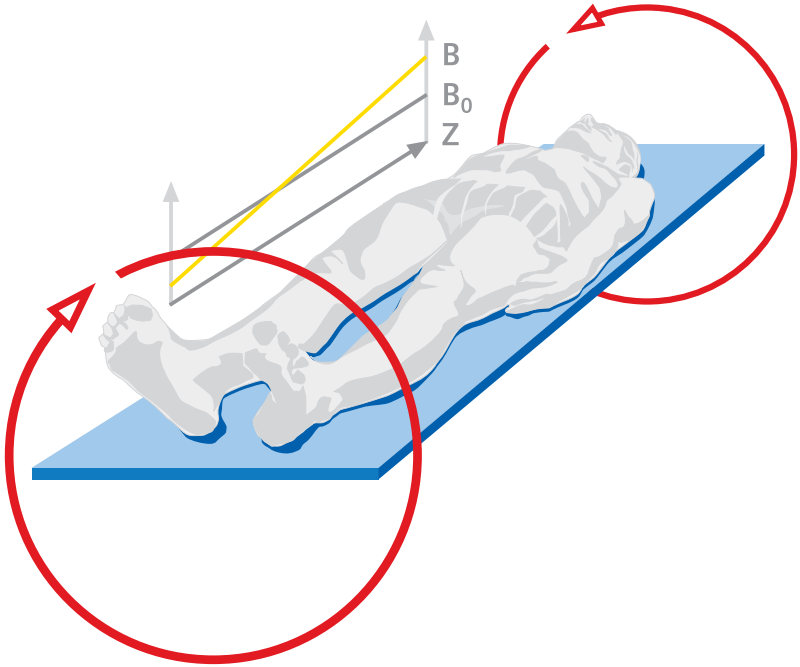
The trick with the gradients

In medical imaging, we would like to take slice images of the human body at specific slice positions.

For this reason we need a method that spatially differentiates MR signals. We can do this in a rather sophisticated manner: by switching gradients.

As you already know: a GRADIENT is a change in magnetic field in a specific direction.





How do we generate a gradient?

A magnetic field is created as soon as an electric current flows through a circular conductor or a coil. When you reverse the direction of current, the orientation of the magnetic field changes as well.

With MR, GRADIENT COILS in pairs are operated in the x, y, and z-direction at the

- *same* current strength but
- of *opposite* polarity however.

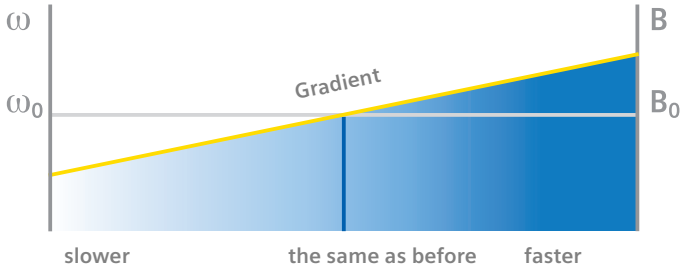
One coil *increases* the static magnetic field, the opposing coil *reduces* it. This means that the magnetic field with its original strength B_0 changes comparable to the incline of a road.

The effect of the gradient

Do you remember the effect of a gradient? Let's review it quickly because it provides us with the basic understanding of MR imaging.

In a normal magnetic field, the field strength is the same everywhere (B_0). For this reason all proton spins show the same spin frequency ω_0 proportional to the field strength. The magnetic resonance is the same everywhere.

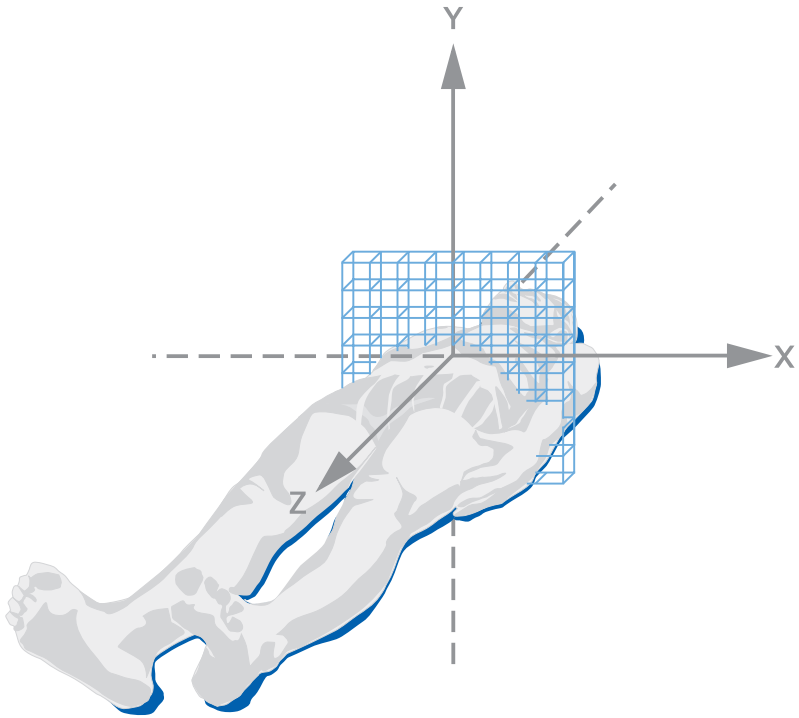
By using a gradient, the magnetic field shows a linear increase. Accordingly the precession of the nuclear spins in this direction varies. They spin more slowly here, or more quickly over there. In sum they show *different* resonance frequencies.

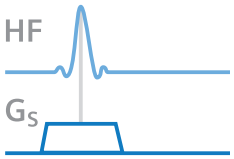


How to determine a slice position

Let's use the following example. Let's pick a slice within the xy-plane, that is vertical to the z-axis. If the patient is in the supine or prone position along the z-axis in the magnet, we are getting a transverse slice.

(Up to now, we have not mentioned one fact: usually the z-axis of an MR system points from the magnet *to the aperture*).

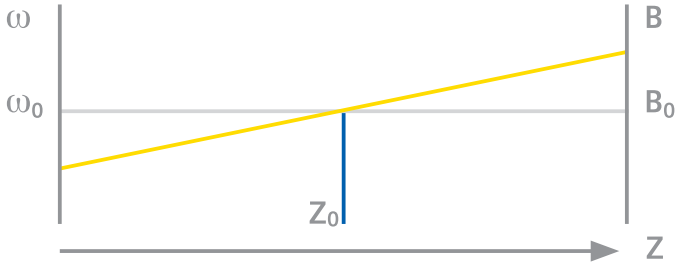




For slice selection, a gradient is switched in the z-direction *simultaneously* to the RF pulse. This gradient is called the SLICE-SELECTION GRADIENT (G_S).

Now the field has its original strength B_0 at one location only z_0. When the RF pulse would possess 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 will get a "slice", but without thickness. The slice would be paper thin and the signal would be too weak, because only a few protons would be stimulated in this thin area. What we need is a certain resolution in the z-direction. And this is what we call SLICE THICKNESS. How do we get that?



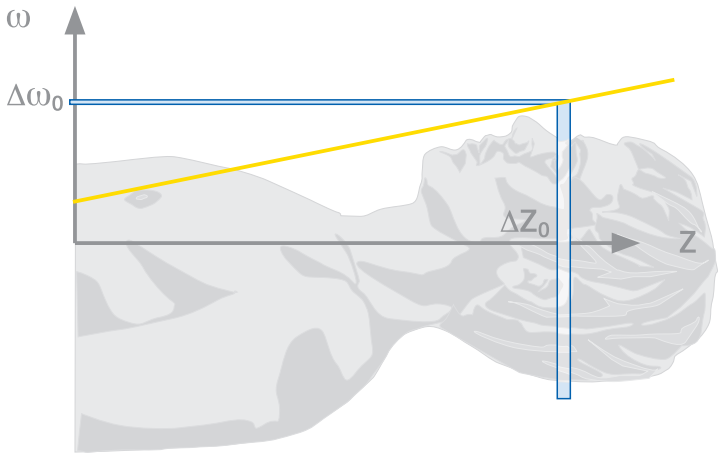
The strength of the homogeneous static magnetic field is B_0. The associated Larmor frequency of the protons is ω_0.

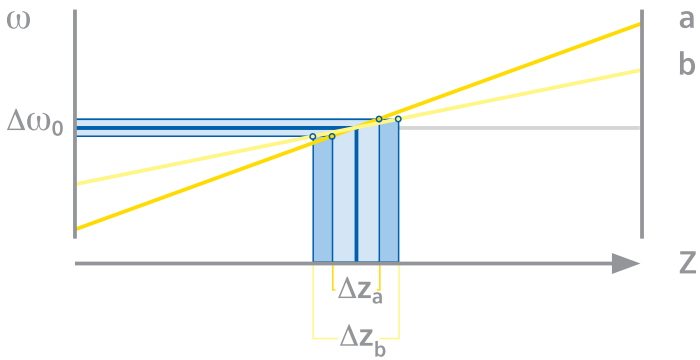
How do we select the slice thickness

The stimulating RF pulse has a certain BANDWIDTH of neighboring frequency about is center frequency ω_0. In this manner it can stimulate the desired spatial area of the slice thickness (Δz_0).

As an alternative: the slice thickness can be modified by holding 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 you select: A SLICE is the *defined resonance area* of a nuclear spin. Outside the slice, the spins are not excited by the RF pulse.
A transverse magnetization (and therefore an MR signal) is generated only within the selected slice.



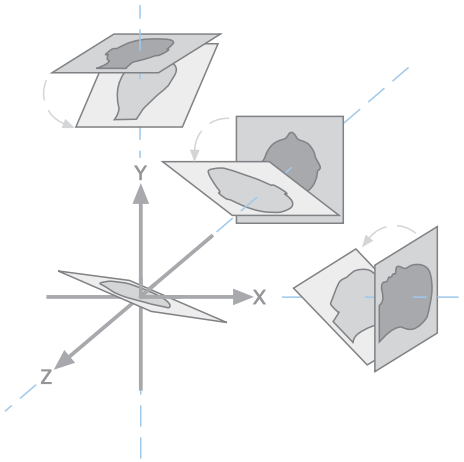


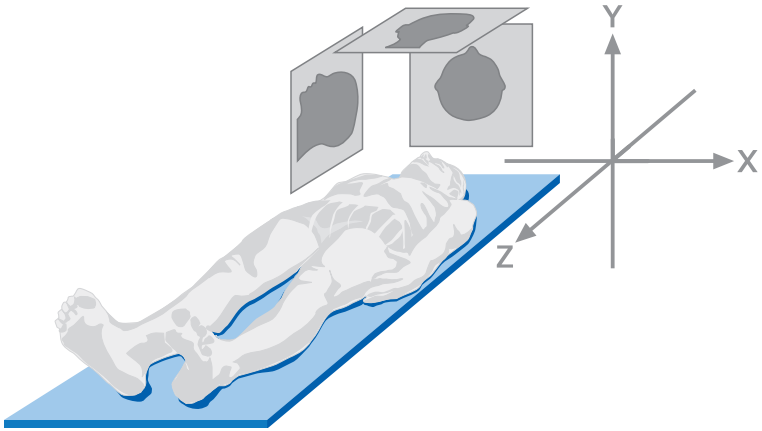
The huge advantage of gradient technology

The gradients allow us to position slice planes at random in MR imaging.

The MR system has three pairs of gradient coils along the spatial axes x, y, and z. For a sagittal slice, we need to switch the *x-gradient*, or the *y-gradient* for a coronal slice.

To obtain OBLIQUE SLICES, we have to simultaneously switch several gradients. 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, we'll switch all three gradients simultaneously.





Review

By switching gradients, we can position random slice planes.

The slice-selection gradient allows us to generate a slice, a spatial region where nuclear spins will resonate.

Outside the slice, the nuclear spins are not affected by the RF pulse.

A stroll through k-space

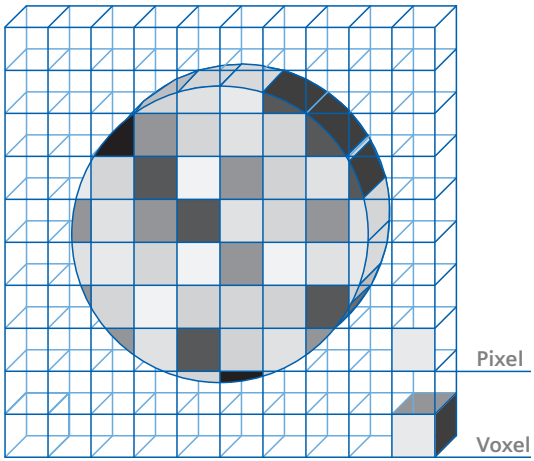
We are now getting to the most interesting section. How do we get an image from the slice? What is important in this case is: the image is *not* directly generated through the measurement procedure. Instead RAW DATA are generated as a first step from the MR signals received. The image is then computed from these raw data. Let's follow this process step-by-step.

The MR image under the magnifying glass

The MR image comprises many individual image elements, known as PIXELS or picture elements. This configuration is known as IMAGE MATRIX. Each pixel in the image matrix has a certain gray value. Seen in total, this gray value matrix results in an image display.

The pixels in the image represent the VOXELS in the slice.

The more pixels are in an image, the more image information is available. This means, the sharper and more detailed is the image. More pixels and/or voxels simply mean a higher RESOLUTION.

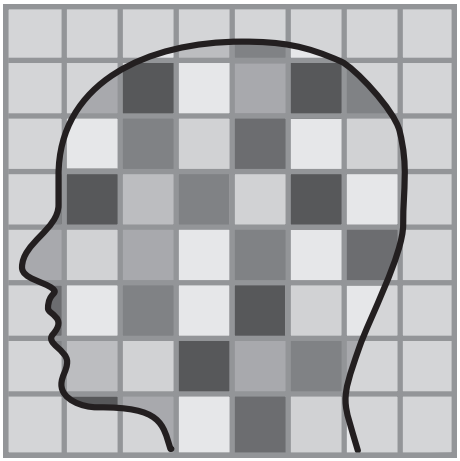


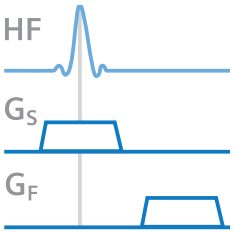
The problem of image generation

Our problem with respect to image generation is how to obtain for each individual voxel in the slice a signal information which generates the gray value of the associated pixel.

Let's assume we want to generate a tomogram in a 256 × 256 MATRIX SIZE. In this case, each row and column have to be differentiated into 256 locations. We use the MR signal to create an image with 256 different values, that is 65,536 voxels.

Again, how do we do that?



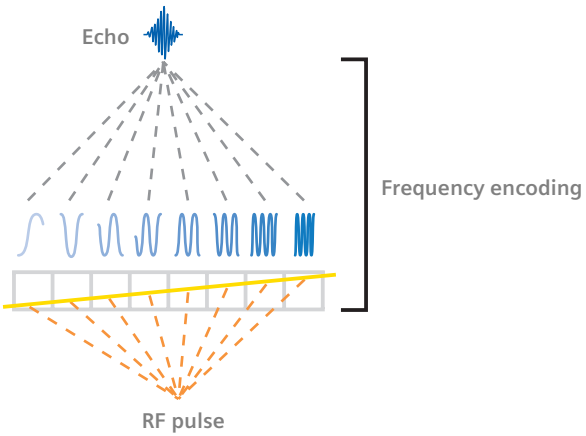


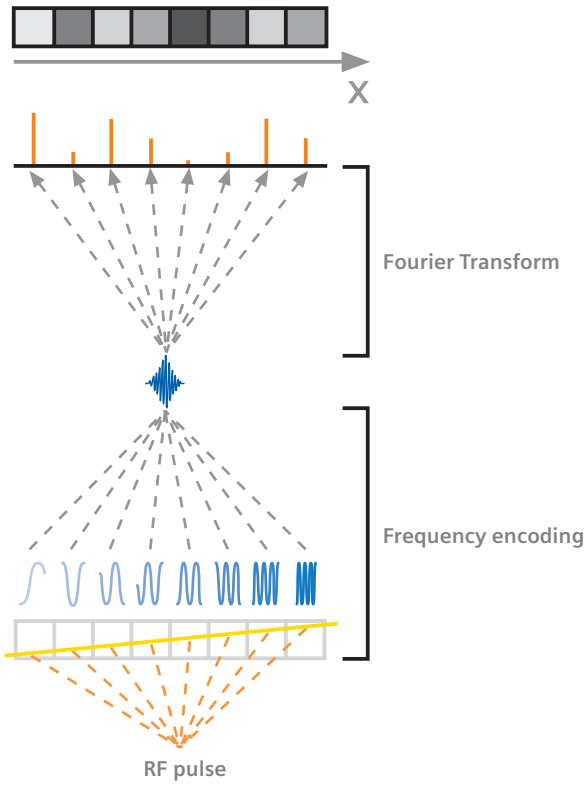
The image of a stripe

For simplicity's sake, we are going to generate only a voxel stripe in place of a 2-dimensional image. The stripe would be running along the x-axis and contain 256 voxels (shortened to 8 in the graphics image).

The signal values can be differentiated as follows. *During* the measurement of the echo, we switch a gradient into the x-direction. What is going to happen? The spin ensemble of the individual voxels precesses 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).

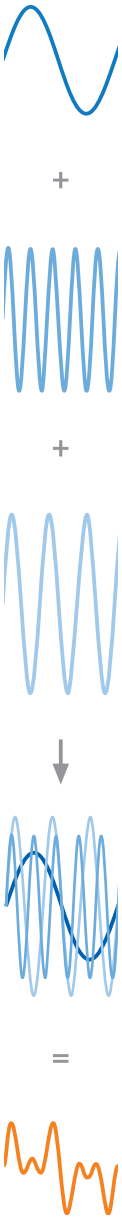
The echo is then a *mixture* of the signals of all excited spins along the x-axis. At a resolution of 256 voxels, the echo includes a mixture of 256 frequencies, a sound that comprises 256 different pitches/tones.
How does this help us?





A multifaceted mathematical procedure—the **Fourier Transform**—allows us to determine the signal contribution of each frequency component (as shown in the graphic by the height of the lines). The individual frequencies are re-allocated to their location of origin on the x-axis. The individual signal obtained determines the gray value of the allocated pixel.

At this point we would have solved or image generation problem if we would limit it to the display of a stripe.



Fourier Transform and signal mixture

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

How should one see a signal mixture? To the left you see three sine curves to be superimposed. The result is a completely new oscillation picture.

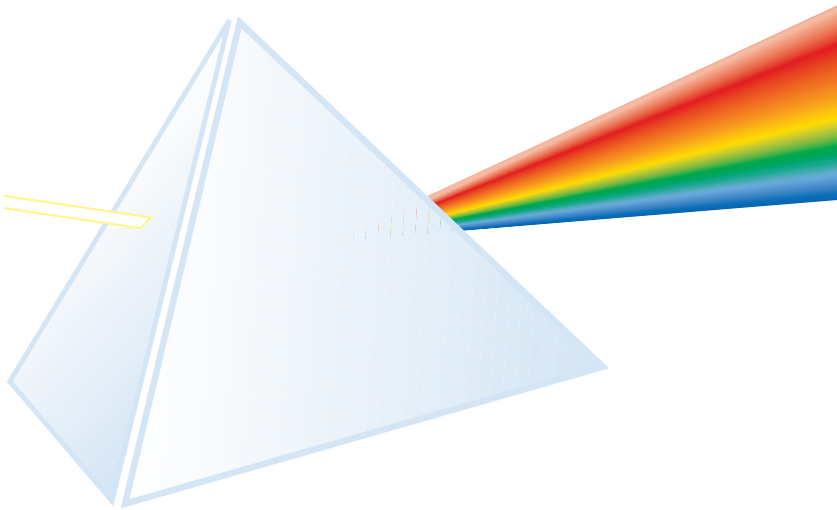
What we can do is to built a given structure from the "building blocks" of sine curves. The more curves we use, the finer the results. The curve shown at the bottom is the result of 32 superimposed sine curves.

What do we understand by Fourier Transform?

Don't get turned off by a mathematical-technical term. Each sound consists of many different pitches, filtered out by your sense of hearing. We would call this ability a natural Fourier Transform.

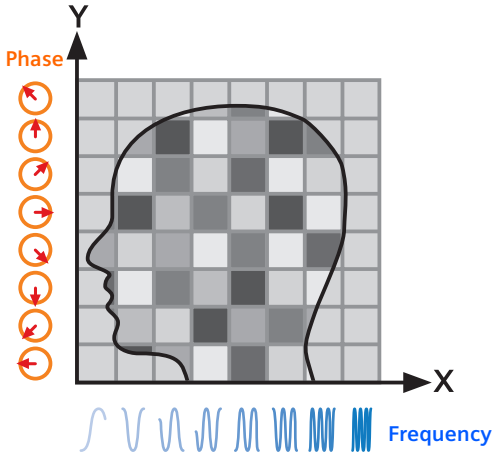
White light is also a mixture of light consisting of different wave lengths or frequencies. A prism acts as a Fourier Transform device in that it aids us in differentiating the various frequencies in white light—in this case the colors of a rainbow.

A Fourier Transform assigns to a structure/signal the individual waves/frequencies which it comprises. This physical dispersion is called SPECTRUM.



From the stripe to the image

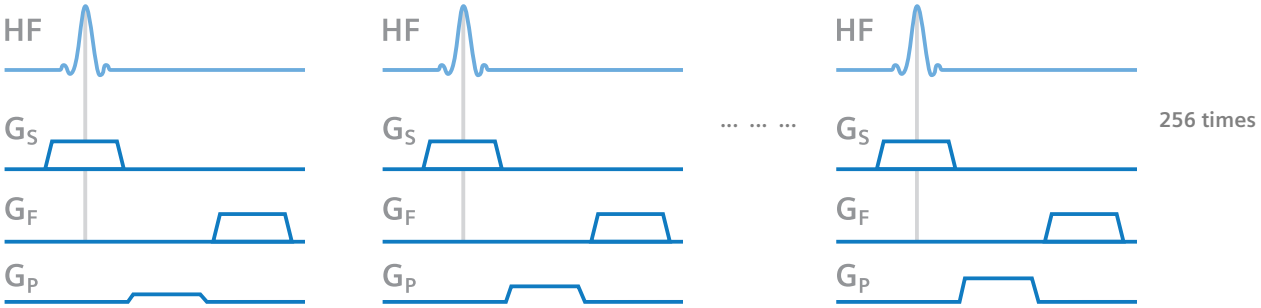
We could not arrive at the idea to use t he same frequency encoding trick in the y-direction to encode a 2-dimensional image. In this case however, two different voxels could have the same frequency and thus could not be differentiated. Obviously we have to choose another path.



During the time *between* the RF pulse and the echo, a gradient into the y-direction is switched briefly. As a result, the spins will precess at different speeds for a short time. After the gradient is switched off, the spins along the y-axis will 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).

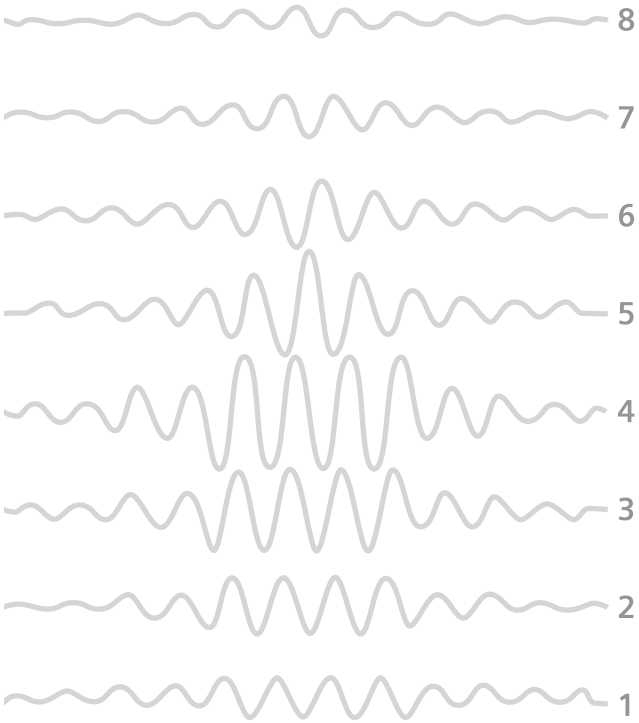
Does this make sense to you?



You can use the Fourier Transform to filter out these phase shifts. There is one slight reservation however. It will only happen when we generate 256 MR signals with different phase encodings for 256 different locations. In other words, this means 256 PHASE ENCODING STEPS. This is the reason why the pulse sequence has to be repeated 256 times for a 256 × 256 matrix.

Thus, a RAW DATA MATRIX is filled line by line with the echoes (shortened to 8 in the graphics image).

This raw data matrix is also known as k-SPACE (a notion used in wave physics).

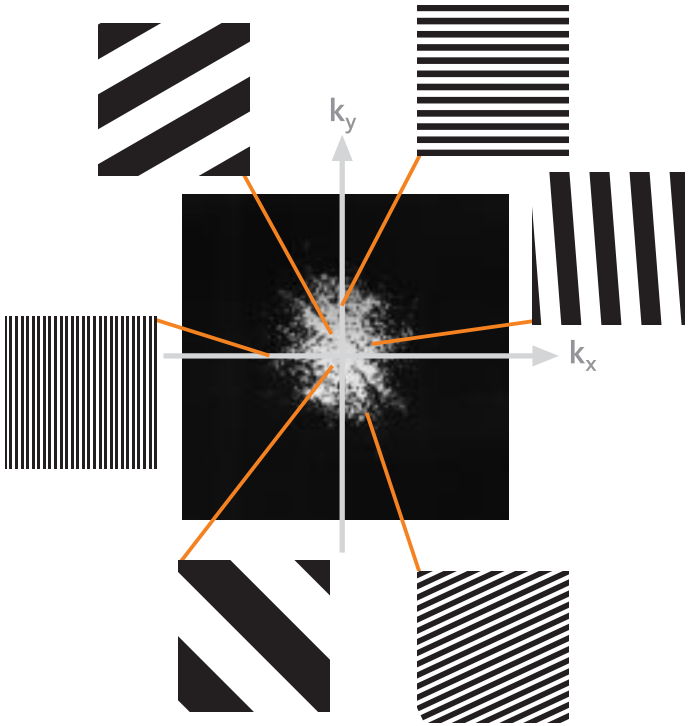


This is how k-space works

Let's take a look at the mysterious k-space. The axes k_x and k_y of the k-space designate so-called SPATIAL FREQUENCIES. What do you think they are?

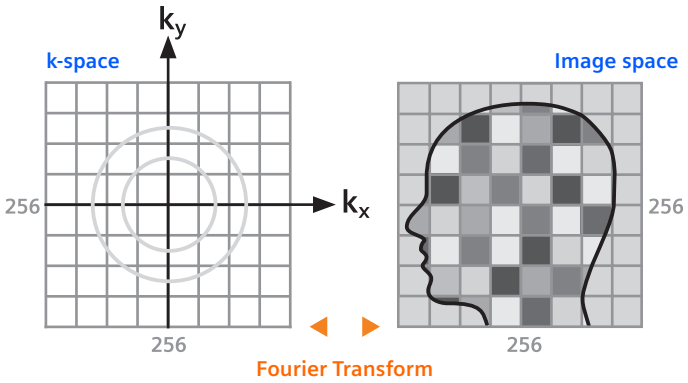
Just like temporal oscillations combine waves of different frequencies, you can compose an image from spatial stripe patterns.

The raw data value in the k-space determines whether and how strong a certain stripe pattern contributes to the image. A rough stripe pattern shows a low spatial frequency (close to the center) a fine stripe pattern shows a high spatial frequency (far from the center).



This is a very simple example to help illustrate our case. By simply superimposing the horizontal and vertical stripe pattern, we generate a complex gray value pattern. You can image that a weighted superimposition of stripe patterns from different spatial frequencies result in a complex image (do you remember the profile built from simple sine waves? refer to page 113).

This is exactly what the 2-DIMENSIONAL FOURIER TRANSFORM does. It uses the raw data values in the k-space for calculating the gray value distribution in the image, that is the weighting of the stripes. Subsequently, it assigns the associated gray value to each pixel.



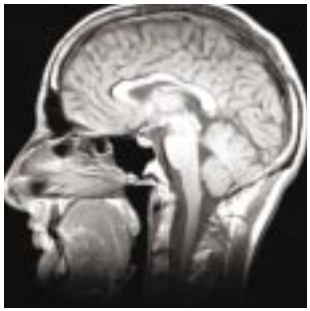
Raw data versus image data

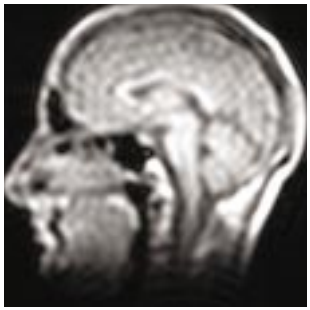
As you saw a dot in the k-space of the raw data does *not* correspond to a pixel in the image, at least not directly.

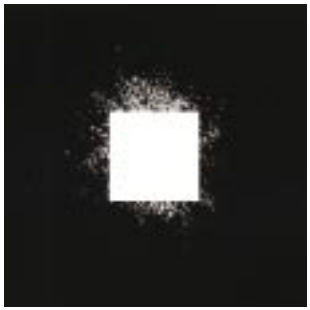
What is true is that each part of the raw data matrix contains information of the whole image—comparable to a hologram.

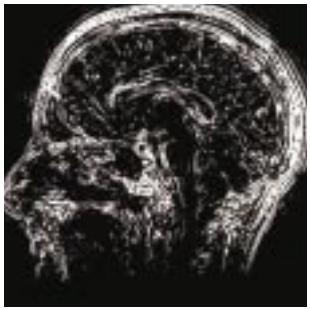
The CENTER RAW DATA determines the rough structure as well as the image contrast.

The RAW DATA ALONG THE MARGIN provide information regarding margins, edge transitions, contours in the image—in short they show finer structures and in the final analysis determine the resolution. They contain almost no information about tissue contrast.









Review

MR imaging technique does not measure the image directly. Instead a scan matrix is filled with raw data.

To localize the individual voxels, the phase-encoding gradient and the frequency-encoding gradient are switched.

The scan matrix acts like a k-space of spatial frequencies. Each spatial frequency corresponds to a specific stripe pattern.

Via a 2-dimensional Fourier Transform, the MR image is computed from the raw data.

Introducing the pulse sequence

Now we have finally gathered all the components for understanding a pulse sequence. The basic run of sequence includes the following: RF stimulation of the spins and slice selection, phase encoding, frequency encoding as well as read-out of the echo.

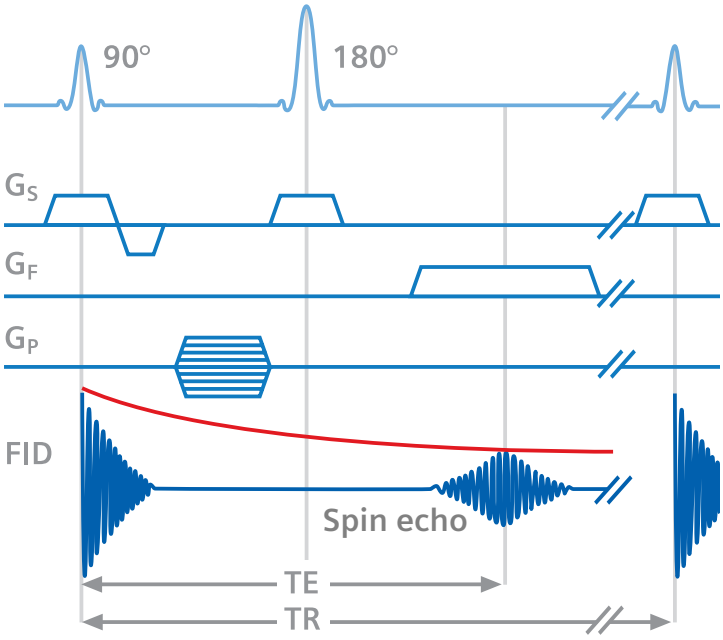
The pulse diagram

Our example involves a spin echo sequence. It consists of a 90 degree pulse, followed by a 180 degree pulse which generates the spin echo in echo time TE.

This pulse sequence is repeated with REPETITION TIME TR as often as the k-space 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 phase-encoding direction.

Scan time = $N_P \times TR$

(N_P: number of phase-encoding steps)

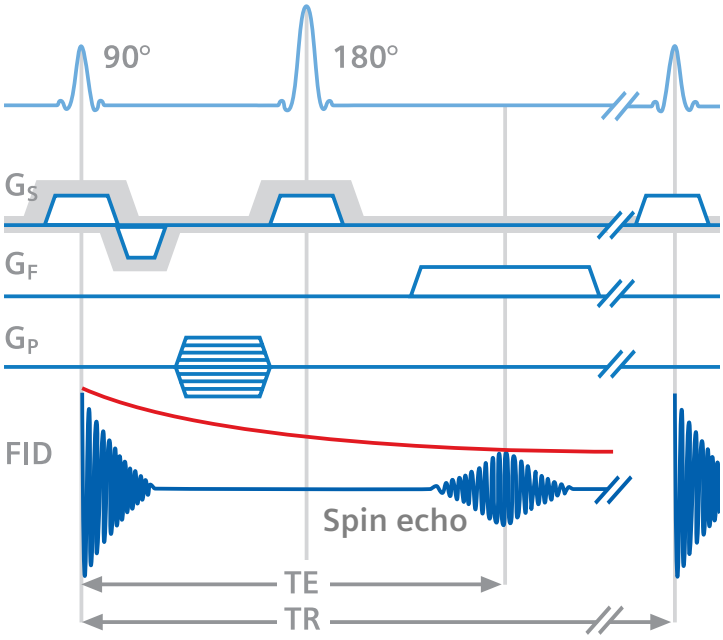


Slice selection

Simultaneous to the 90 degree pulse, the SLICE-SELECTION GRADIENT G_S is switched (bar pointing up). This is how the slice is selected.

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

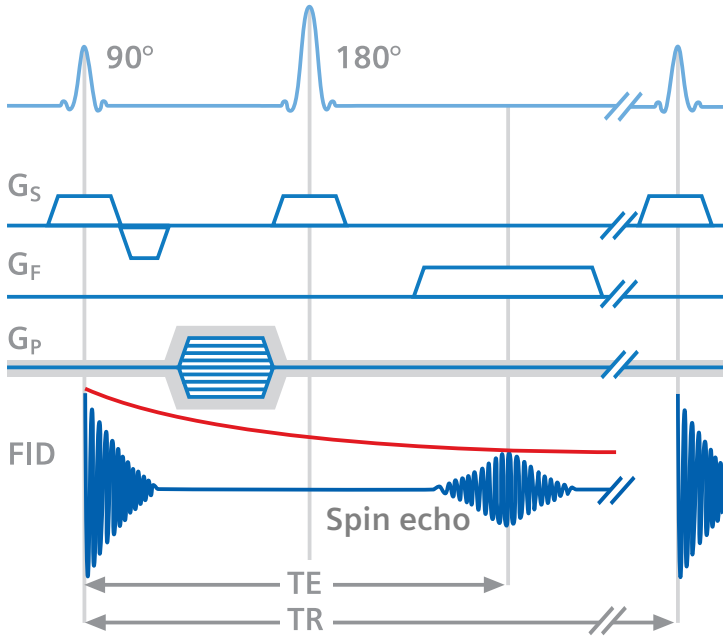
During the 180 degree pulse, the slice-selection gradient is switched again so that the 180 degree pulse only affects the spins of the previously stimulated slice.



Phase encoding

The PHASE-ENCODING GRADIENT G_P is turned on briefly between slice selection and spin echo. The phase-encoding gradient superimposes a different phase on the spins.
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.

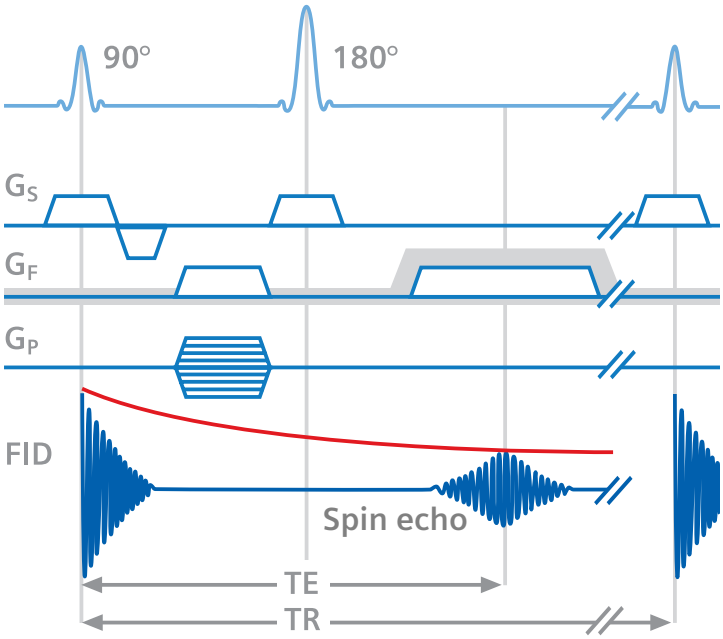


Frequency encoding

During the spin echo, the frequency-encoding gradient G_F (second long bar) is in effect. Since the spin echo is read out during this time, this gradient is also known as the READOUT GRADIENT.

If we apply nothing else but 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 an additional gradient.

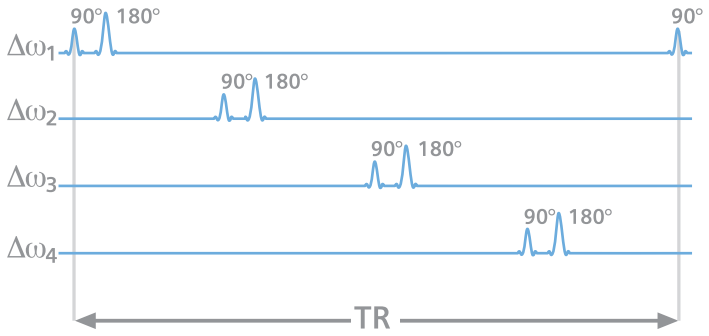
Prior to readout, the spins are dephased by a gradient of opposite polarity and half the duration of the readout gradient (dephasing gradient). This tricks lets the readout gradient rephase the spins again in a way which causes the spins in the center of the readout interval to be in phase again at the time of the maximum spin echo. When, as our example shows, we turn on the dephasing gradient before the 180 degree pulse, this gradient has the same polarity as the readout gradient. Why? The 180 degree pulse reverses the phase of the spins.

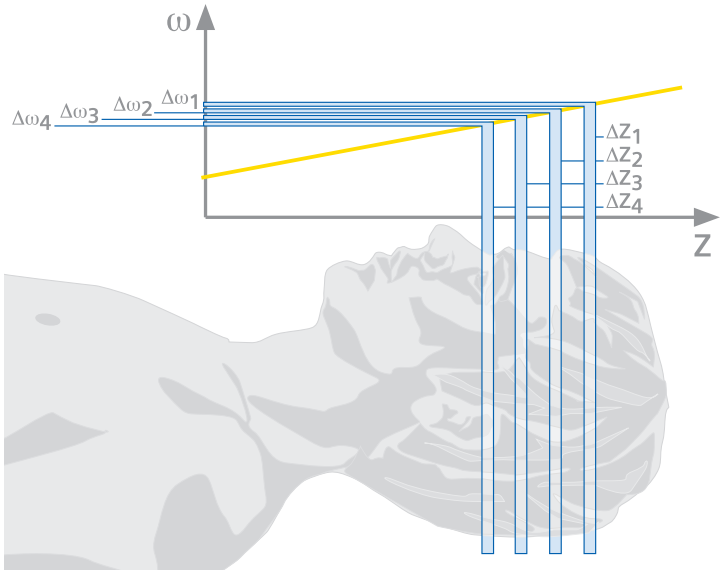


We are measuring several slices at once

Echo time TE is always considerably shorter than repetition time TR. During the time interval between reading out the last echo and the next RF pulse, we can excite additional slices (e.g. z_1 through z_4). The result will be a MULTISLICE SEQUENCE.

This method provides us with all the slices necessary for examining a particular area or region during a measurement.



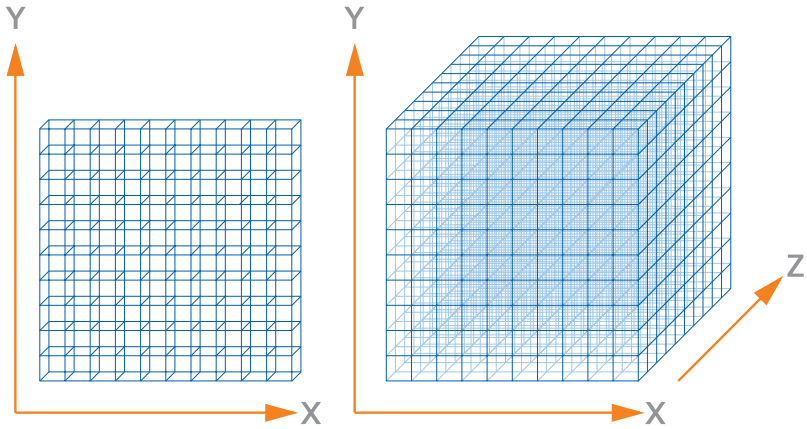


How to create 3-D data

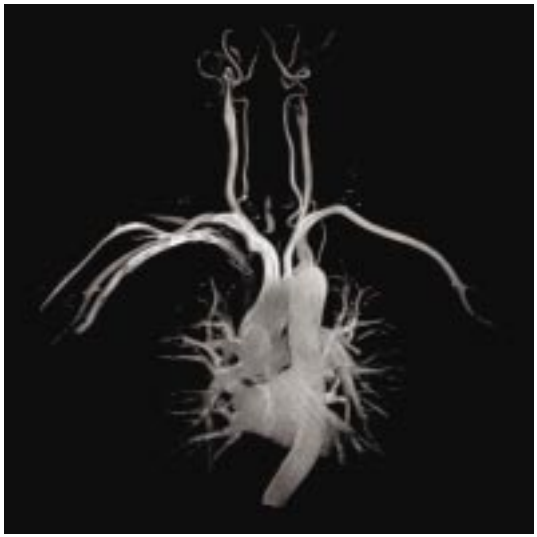
Faster sequences such as e.g. gradient echo sequences provide a considerable advantage: they allow us to create 3-D data sets because of the short repetition time. These 3-D data sets are used for reconstructing 3-dimensional displays.

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

Through the additional phase encoding perpendicular to the image plane as well as contiguous images we obtain information regarding the spatial volume (SLAB). The planes of this volume are known as PARTITIONS.



From the data set generated during the 3-D measurement, the post-processing software can create spatial views.

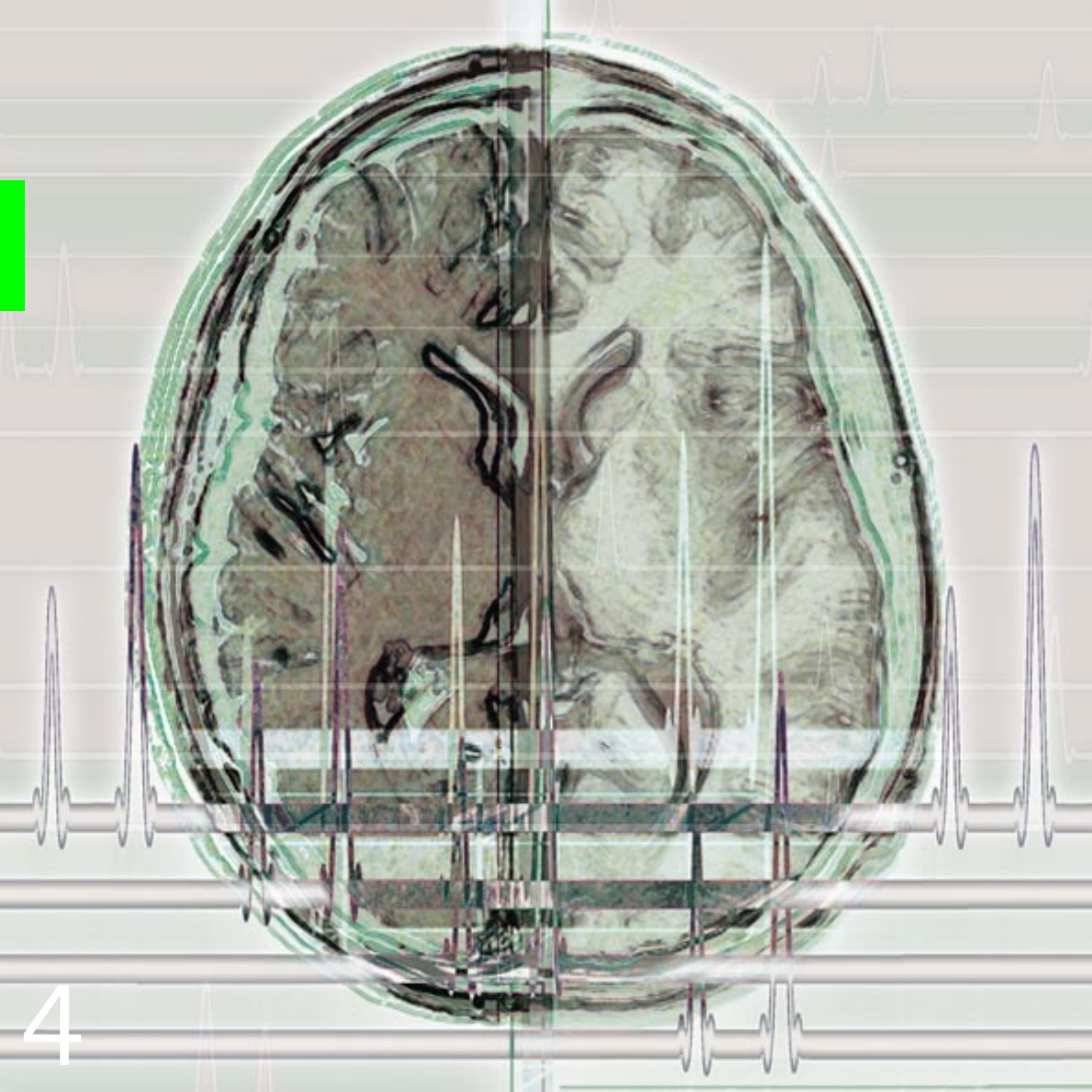


The principle of MR imaging

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

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

With the help of 2-dimensional Fourier Transform, the MR system reconstructs the MR image from the raw data measured.



The wide range of contrasts

The quality of the image contrast determines the diagnostic relevance of a medical image. MR imaging is unique in its possibilities of controlling image contrast and therefore expanding the diagnostic range.

The art of MR application lies in the choice of pulse sequences and the combination of acquisition parameters.
This chapter shows you the most important pulse sequences and types of contrast.

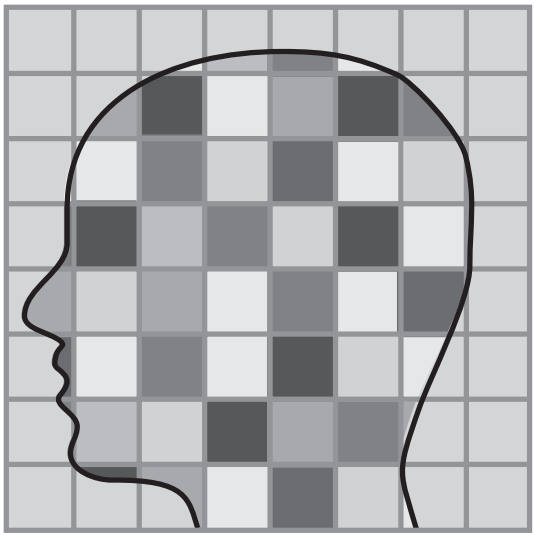
The attachment of this chapter provides you with a brief introduction to MR spectroscopy.

Spin echoes and contrast weighting

Using a spin echo sequence as an example, we are able to demonstrate the three most important types of contrast in MR imaging: T_1 contrast, T_2 contrast, as well as proton density contrast. All three contrast types contribute more or less to the image contrast, but it is usually only one that determines the contrast as such. This enhancement of a contrast type is usually called WEIGHTING.

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 from the proton density in the respective voxel: the greater the number of protons contributing to the magnetization, the stronger the signal.

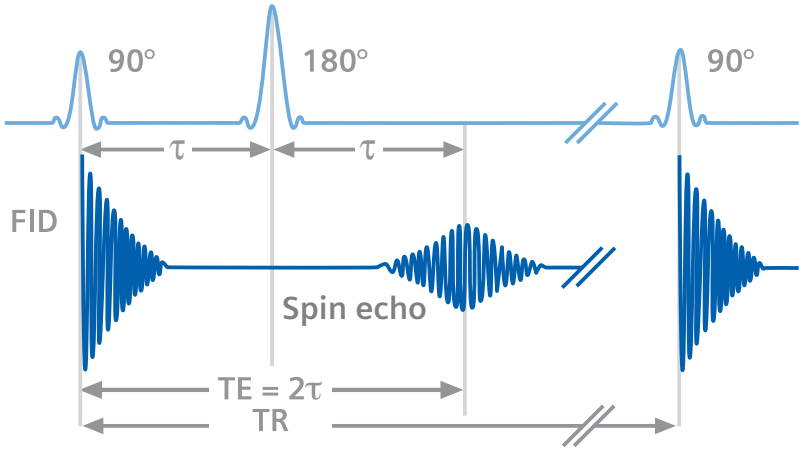
But even more important for medical diagnostics is the effect of the two relaxation constants T_1 and T_2 on the image contrast.

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 = 2τ.

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

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

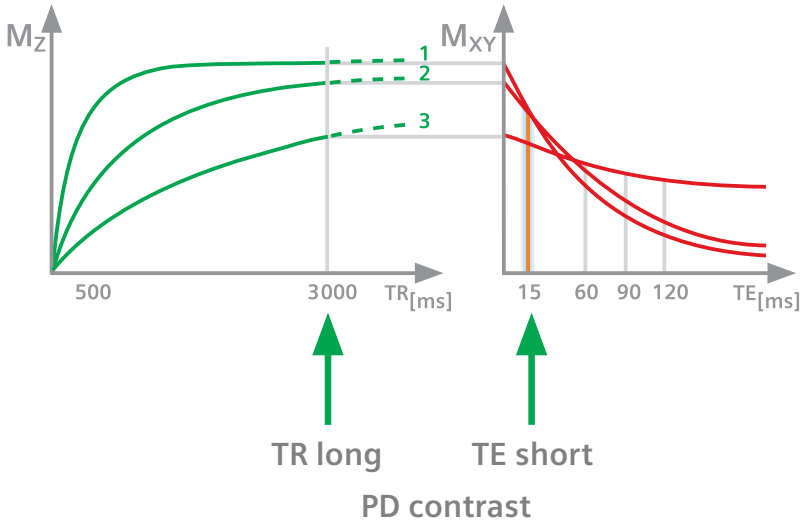


Proton density contrast

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.

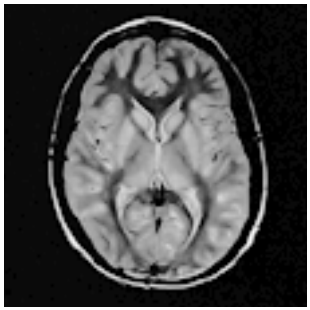
By means of a repeated 90 degree pulse after repetition time TR, the actual longitudinal magnetizations convert into transverse magnetizations M_{xy} and generate signals of different strength.



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

In actual application, the TR of a spin echo sequence is rarely longer than 2–3 seconds. However, this also means that tissue types with longer T_1 constants, e.g. CSF, have not recovered after this time period.



Proton density contrast:
TR long (2,500 ms)
TE short (15 ms)

The larger the proton density of a tissue type, the brighter it appears in the PD image.

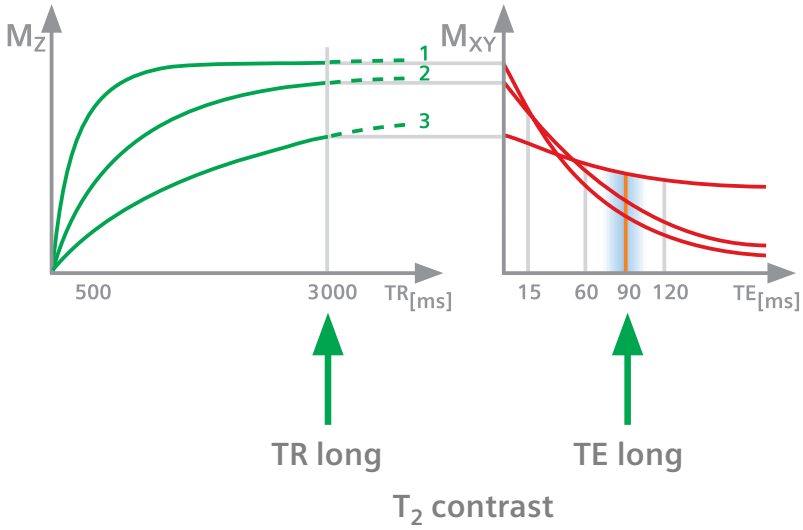
PROTON DENSITY CONTRAST
TR long—TE short

T_2 contrast

Let's 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 begin to intersect. The proton density contrast is lost. At 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.

At increasing echo times, 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 has decayed to a level where the signal of some tissue types disappears in the unavoidable signal noise.

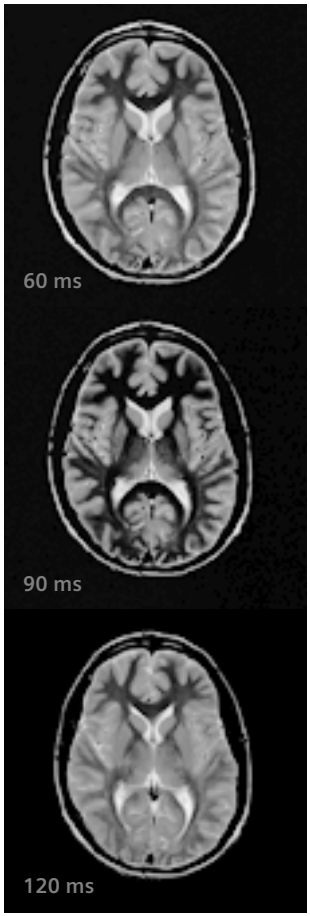


Image comparison with respect to T_2 contrast:
TR long (2,500 ms)
TE is increasing

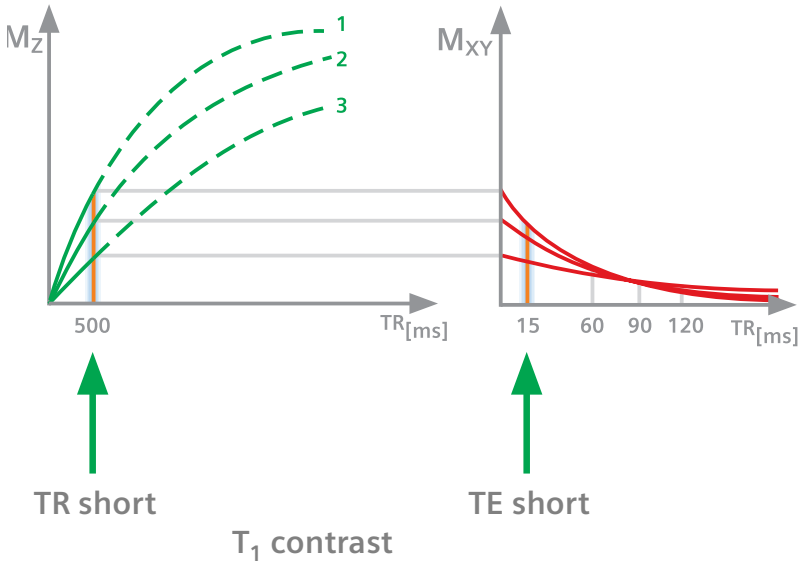
CSF with a long T_2 appears bright in a T_2-weighted image.

T_2 CONTRAST
TR long—TE long

T_1 contrast

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 decreases rapidly with the increasing echo time. For this reason, we have to select the *shortest* possible echo time TE.

A short TR cancels the effect of the proton densities, the 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.



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.

T_1 CONTRAST
TR short—TE short

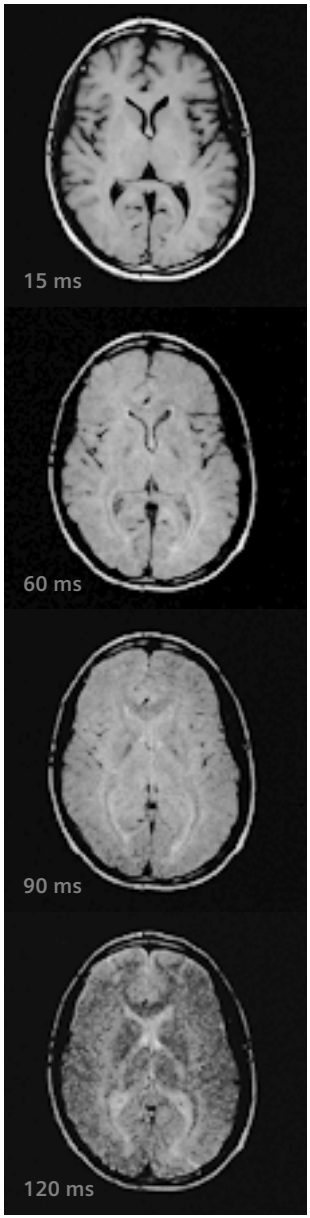


Image comparison with respect to T_1 contrast:
TR short (500 ms)
TE is increasing

CSF with a long T_1 appears dark in a T_1-weighted image.

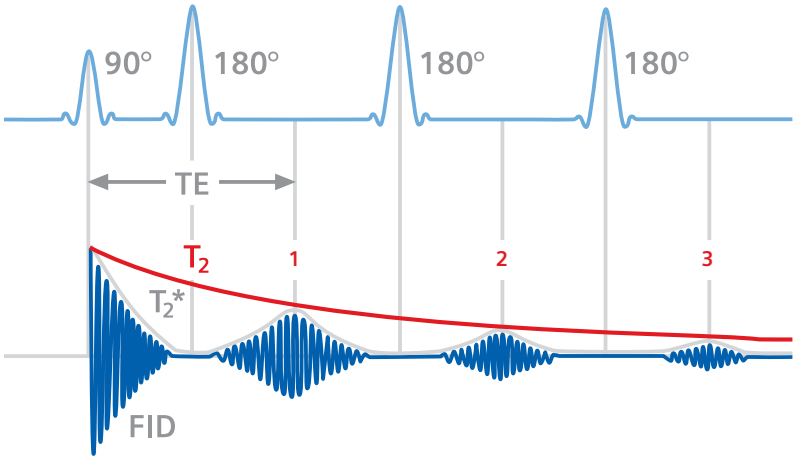
The optimal TR corresponds approximately to the average T_1 constant of the tissue type to be displayed. This means between 400 ms and 600 ms for 1.0 to 1.5 tesla.

Measuring multiple echoes

We can create two or more spin echoes with a MULTI-ECHO SEQUENCE. The signal strength of the echo decreases with T_2 relaxation. This drop in signal allows us to compute a pure T_2 IMAGE from the data, without T_1 portions.

Similarly, we can compute a pure T_1 IMAGE from the signal strength of several spin echo measurements with different repetition times TR but the same short echo time TE.

With a DOUBLE-ECHO SEQUENCE (e.g. TE1 = 15 ms and TE2 = 90 ms) we obtain a proton density image as well as a T_2-weighted image from a single measurement.

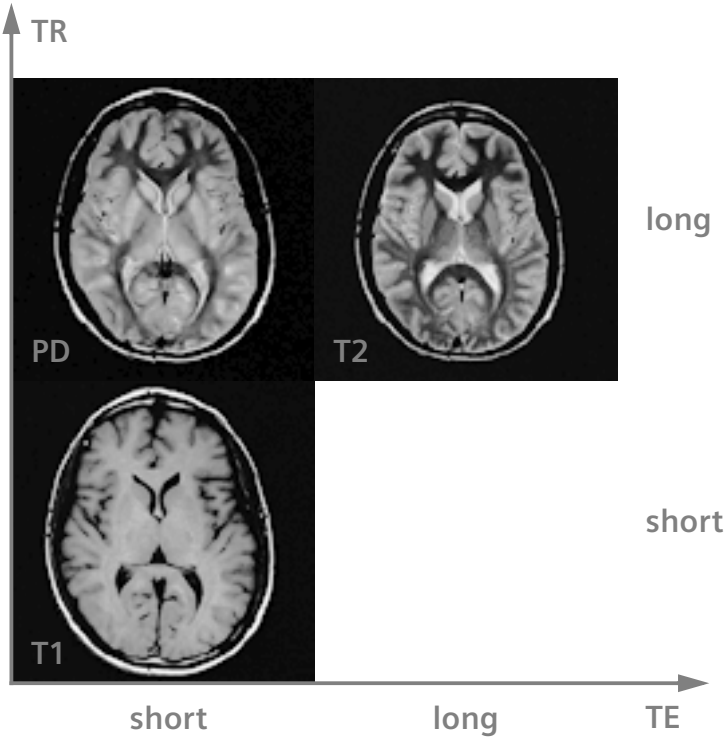


Review

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)

With 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*.



Contrast with Inversion Recovery

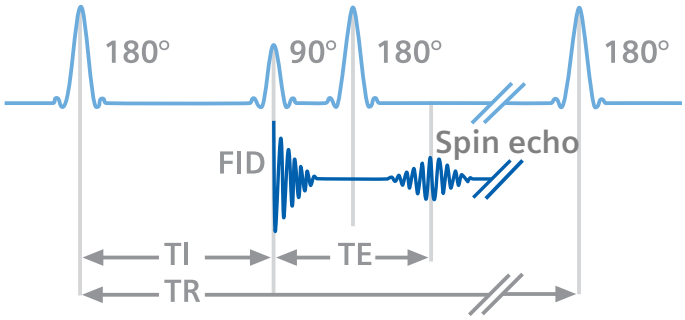
The inversion recovery sequence is a spin echo sequence with a preceding 180 degree pulse. In MR technology, PREPARATION PULSES frequently precede the actual sequence. In this context, our concern is how to manipulate image contrast.

First inversion, then recovery

The INVERSION RECOVERY SEQUENCE (IR) 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. The 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 stimulating 90 degree pulse converts the current longitudinal magnetization into transverse magnetization.

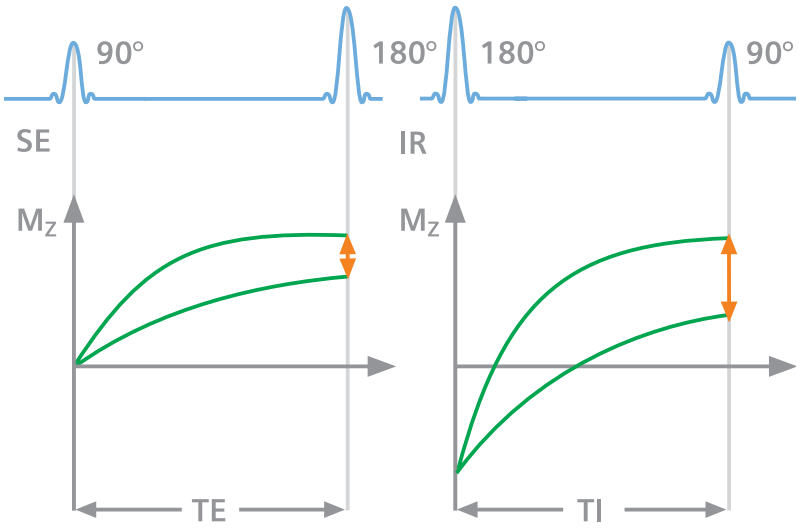


Strong T_1 contrast

While the spin echo sequence provides for 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 gives improved dispersion of these TI curves leading to better T_1 contrast. By selecting a suitable inversion time TI, the contrast is optimized.

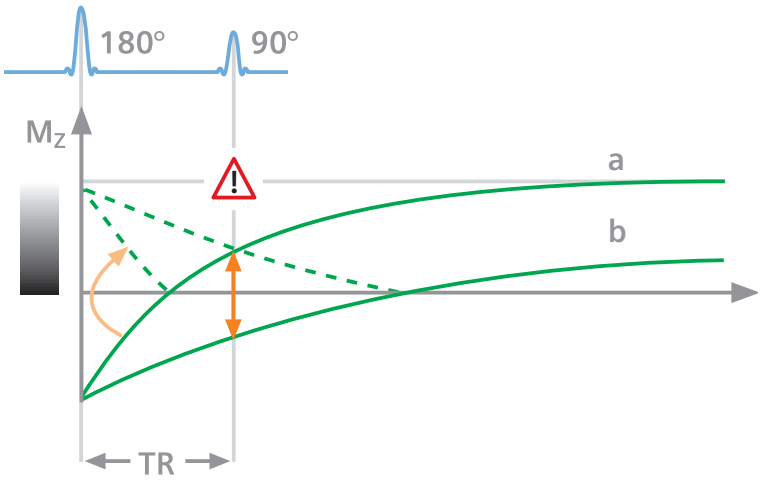
We can use IR sequences to display the most minute T_1 contrasts, for example, in the brain of a newborn. The disadvantages are the longer measurement time. Depending on TI, fewer slices are measured than with a T1-weighted spin echo technique.



Gray on gray and the zero signal

Let's 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-crossing, 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.

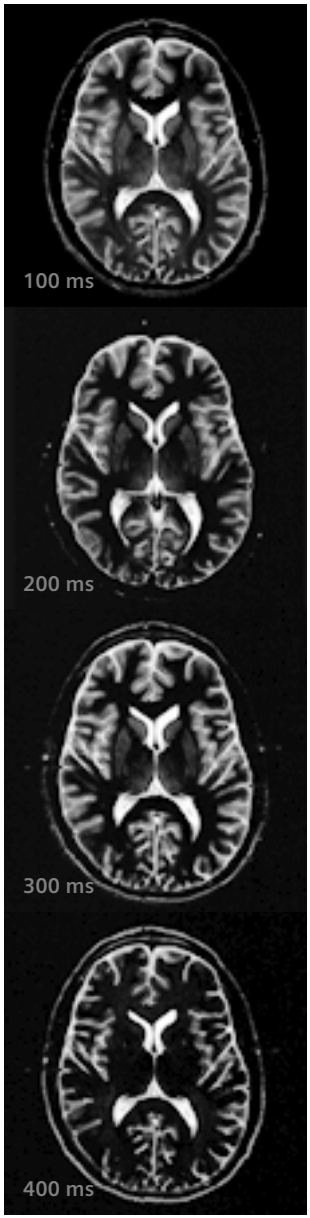


Image comparison with respect to contrast with inversion recovery: TI is increasing

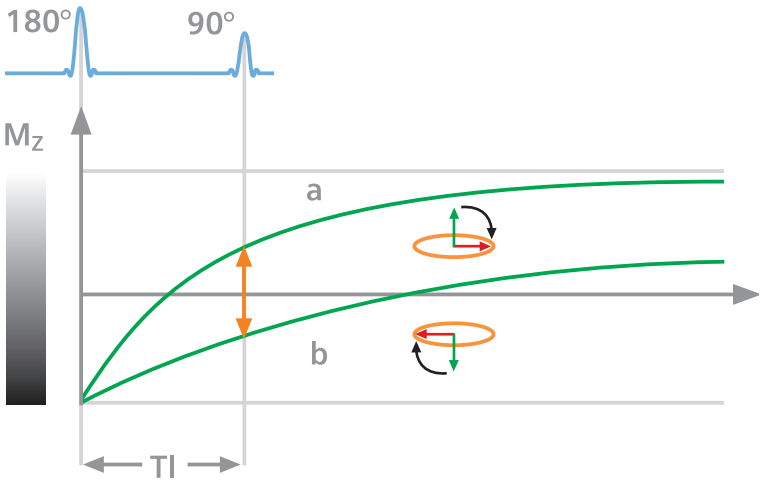
The signal of white matter decreases as the inversion time TI increases and goes through the zero-crossing at TI = 300 ms.

At TI = 400 ms, the signal of gray matter (with longer T_1) has reached zero-crossing, while the signal for white matter is increasing again.

T_1 contrast all the way

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 a 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.

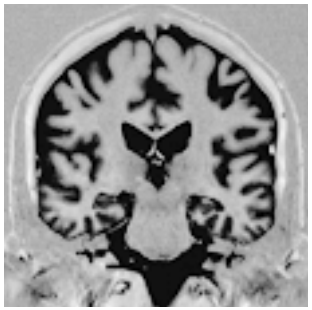


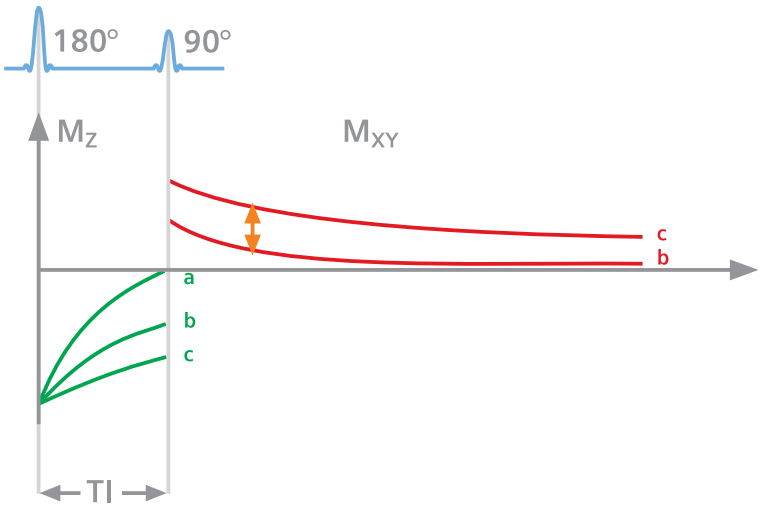
Image background, usually black, is shown in mid-range gray when using phase-sensitive reconstruction.

Additive T_1 and T_2 contrast

To reiterate spin echo imaging: tissue with a longer T_1 appears darker in the image, tissue with longer T_2 appears brighter. In short, T_1 and T_2 oppose one another.

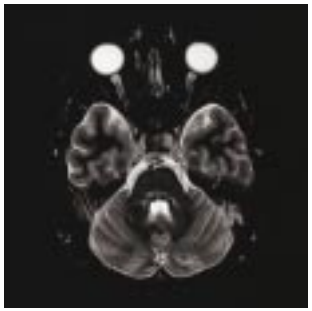
When using a short inversion time, the inversion recovery technique obtains a rather interesting contrast: additive T_1 and T_2 weighting (this sequence is known as STIR = Short TI Inversion Recovery).

In this case, tissue with a long T_1 (b, c) shows a negative longitudinal magnetization. After the 90 degree excitation pulse, the tissue generates stronger signals (T_1 portion). With longer echo times, the contrast is further enhanced (T_2 portion). T_1 and T_2 effects work in the same direction.



Fat appears as very bright in a T_1-weighted image. The results are frequent blooming and motion artifacts.

Ideally, we would select a TI such that fat having the shortest T_1 would have just reached the zero-crossing of the longitudinal magnetization (a).
TI would have to be 0.69 T_1. As a result, the fat signal would be suppressed (TI = 180 ms at 1.5 tesla and TI = 160 ms at 1.0 tesla).



STIR image:
The fat signal is suppressed in the area of the orbits. The optical nerve is clearly delineated.

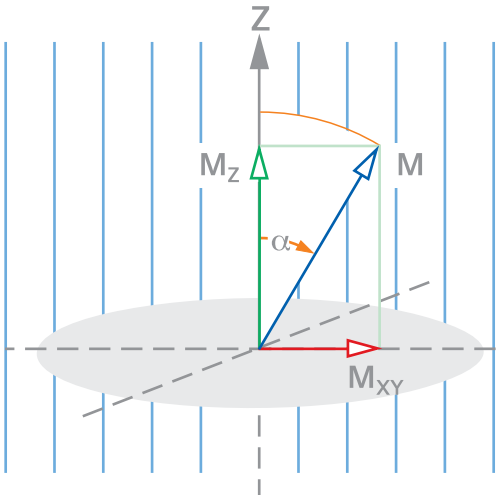
Contrast with gradient echoes

The more you reduce the repetition time TR of a spin echo sequence, the less time remains for T_1 relaxation: the spin echoes are growing weak. However, by using a flip angle of less than 90 degrees, you are able to increase the MR signal and shorten the measurement time as well. For this purpose, you are using gradient echoes.

Shortening the repetition time without signal loss

What happens when the flip angle α of a pulse sequence is less than 90 degrees? In this case, we do not have the effect of the entire available magnetization 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**, instead it continues to have a reduced magnitude M_z.

For example, an RF pulse with a flip angle of 20 degrees already generates a sufficiently high transverse magnetization of 34 % of its maximum value. The remaining longitudinal magnetization reaches in this case 94 % of its maximum value. This allows for very short repetition times and a greatly reduced measurement time.

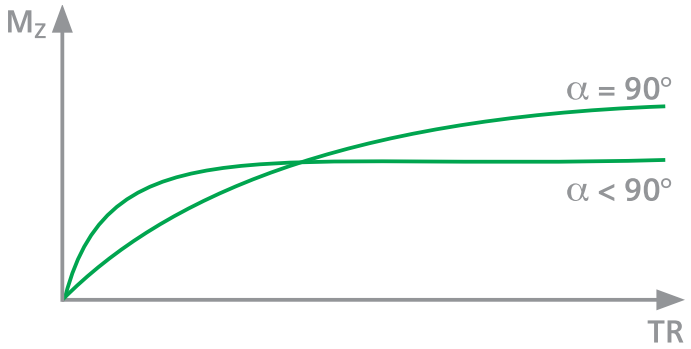


At the next pulse, a high longitudinal magnetization is available again. At very short repetition times (less than T_1), a 20 degree pulse will generate a stronger MR signal than a 90 degree pulse.

Optimal flip 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.



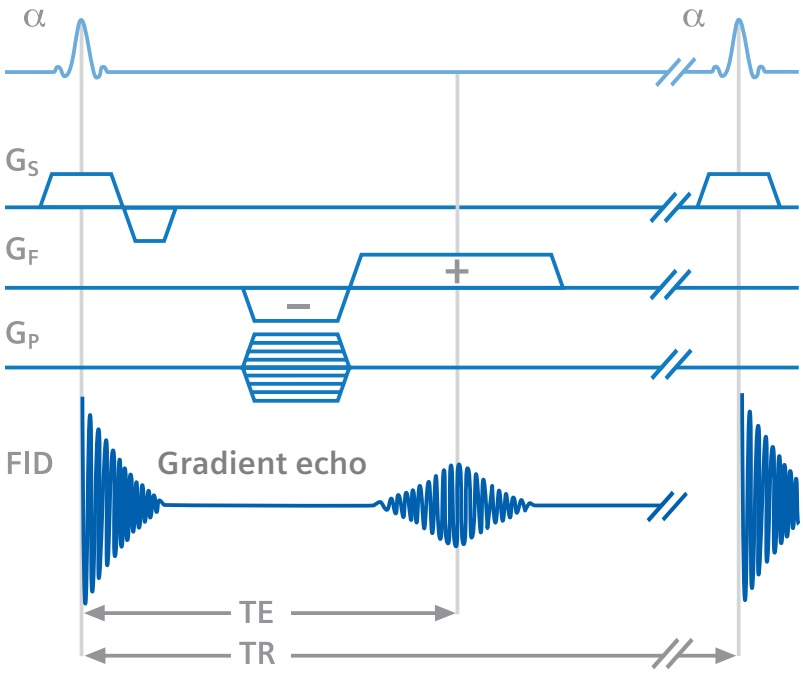
Destroying the transverse magnetization (FLASH) ...

At a very short repetition time TR, we have a residual transverse magnetization prior to each emitted α-pulse.

The FLASH sequence utilizes the steady state of the longitudinal magnetization. The remaining transverse magnetization is destroyed by the phase shifting of the repeated α-pulse.

The FLASH sequence utilizes the steady state of the longitudinal magnetization. The remaining transverse magnetization is destroyed by the strong gradient pulses prior to the repeated α-pulse.

FLASH is the abbreviation for *Fast Low Angle Shot*.



The following types of contrast can be generated with a gradient echo sequence rather than with a spin echo technique and are very complex. The contrasts generated with a FLASH sequence are as follows.

- 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°)

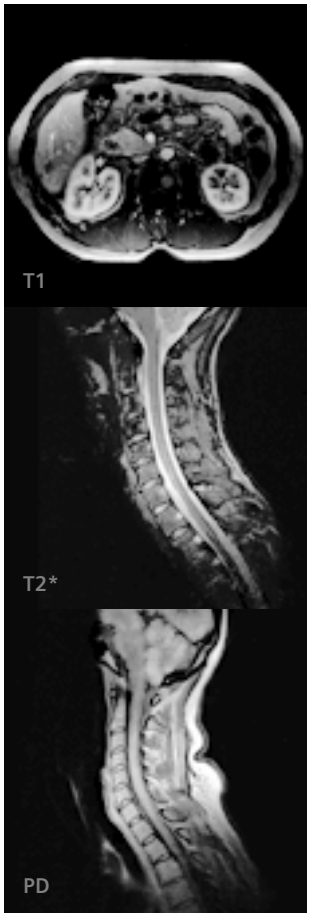
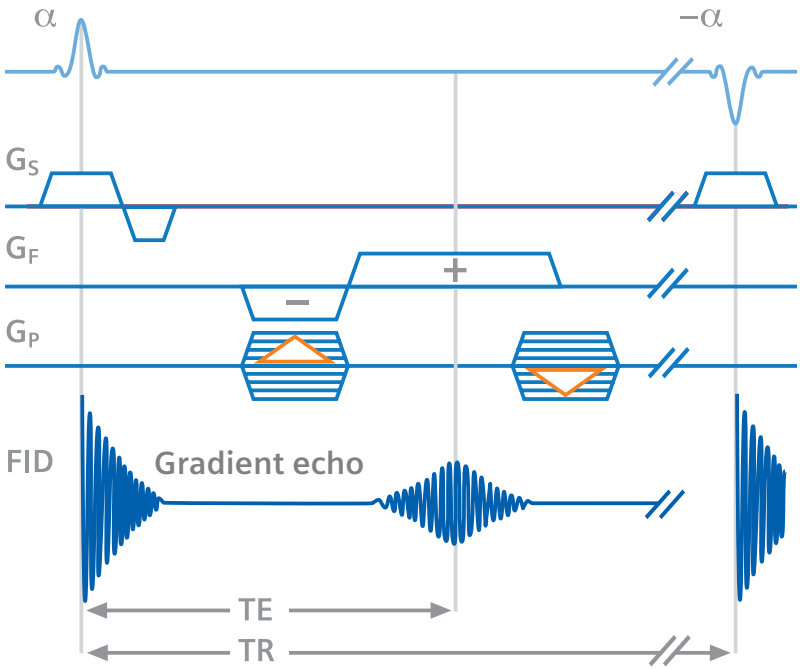


Image comparison showing FLASH contrasts

... or utilizing the transverse magnetization (FISP)

The FISP sequence uses the steady state of the residual transverse 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.

FISP is the abbreviation for *Fast Imaging with Steady-state Precession*.

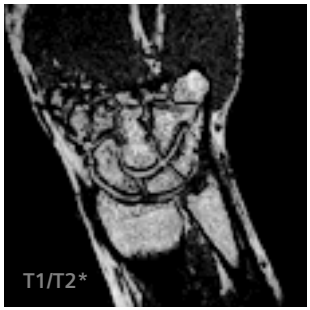


The negative α-pulse (−α) shows that the magnetization is flipped in the opposite direction after each repetition time TR.

The longitudinal magnetization depends on T_1, the transverse magnetization depends on T_2*. The contrast obtained with FISP is a function of the ratio of T_1 to T_2*. It is more or less independent of TR.

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

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



Contrast with 3-D FISP

Attachment: A brief chat regarding MR spectroscopy

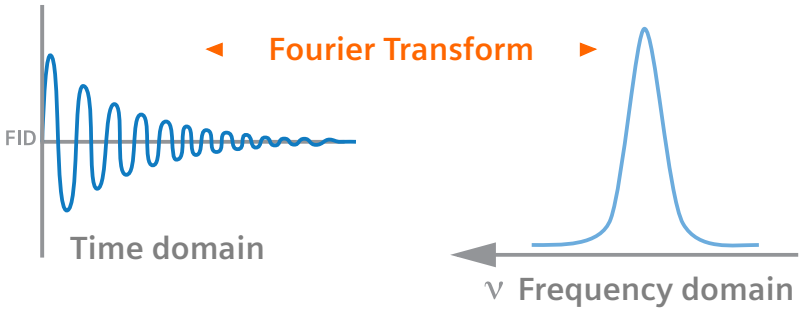
At the end of this chapter about contrasts, we would like to briefly describe another, older MR technique: MR spectroscopy—which is now in clinical use. We will limit ourselves to the simplest method, the single volume method for hydrogen protons (Single Voxel Spectroscopy or SVS).

From the FID to the peak

In MR spectroscopy as in MR imaging, the MR signal is measured as a function of time: the FID is a rapidly decreasing RF oscillation. Echo signals are used as well in addition to the FID.

By using a single Fourier Transform, this oscillation is converted into its frequency components known as the SPECTRUM.

This transformation is a conversion of the signal from the TIME DOMAIN into the FREQUENCY DOMAIN.



If the signal has only one frequency (sine oscillation), the associated spectrum consists of a single RESONANCE LINE only of the associated frequency. Since the signal decays, the resonance line changes into a PEAK of finite width.

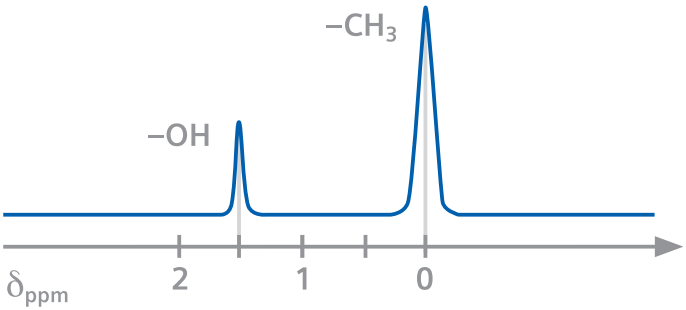
The peak represents the resonance frequency in the voxel measured. What is interesting is that the area under the peak is proportional to the number of signal-emitting nuclei (in this case the proton density).

The chemical shift

In almost all biomolecules, hydrogen atoms are bound in different positions of various molecules. Different positions mean different chemical and therefore quite frequently different magnetic environments. The local magnetic field is reduced and/or increased and the resonance frequencies of the bound protons are somewhat lower or higher than the typical Larmor frequency. For this reason, the nuclei of a molecule can yield *multiple* resonance lines.

This change in resonance frequencies is known as CHEMICAL SHIFT, because they show a shift of the associated resonance lines in the measured spectrum.

Because of the chemical shift, we are able to differentiate molecular components, molecules and substances.

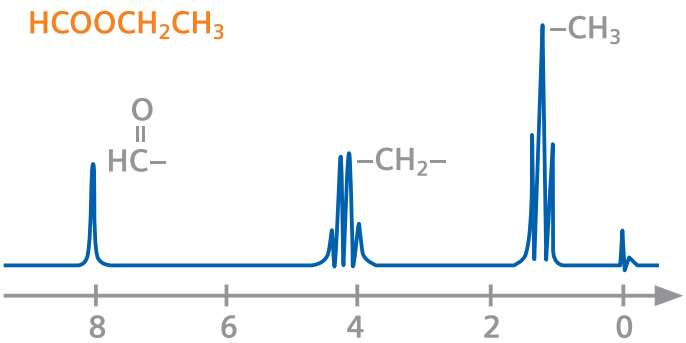


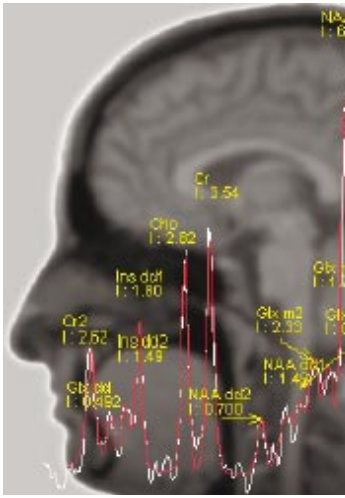
Example Methanol (CH_3OH): The ratio of the peak areas is 3:1. As a result the peaks can be either associated with the hydroxyl group (OH) or with the 3 equivalent hydrogen atoms of the methyl group (CH_3).

The degree of chemical shift can be expressed in δ_{ppm} (ppm = parts per million). $\delta_{ppm} = -1.5$ means the frequency of the OH group has been reduced by 1.5 millionth (this is by 60 kHz at a 40 MHz Larmor frequency).

The fine splitting of the resonance lines

Not all nuclei provide simple resonance lines (*singlets*). Some nuclei show a characteristic splitting of the lines, such as triplets or quartets. The reasons for this are the reciprocal magnetic effects of the nuclei, the so-called SPIN-SPIN COUPLING.





In practical application, the comparison of spectra does not involve absolute peak areas, but rather the relative signal intensities. They are used to compare spectra in healthy tissue with spectra in pathological tissue.

Fast image generation

Ultra-fast imaging provides us today with slice acquisition times in the sub-second range. MR imaging has been greatly accelerated by optimizing the known spin echo and gradient echo techniques. One of the preferred methods is to more quickly fill existing acquisition matrices with echoes than with conventional technologies. On the next pages, we will show you two typical representations of this method: TurboSE and EPI.

Parallel acquisition technologies are both new and superior. They *spatially* optimize the filling of matrices by simultaneously using MR signals from several coil elements.

Turbo measurements with Turbo spin echoes

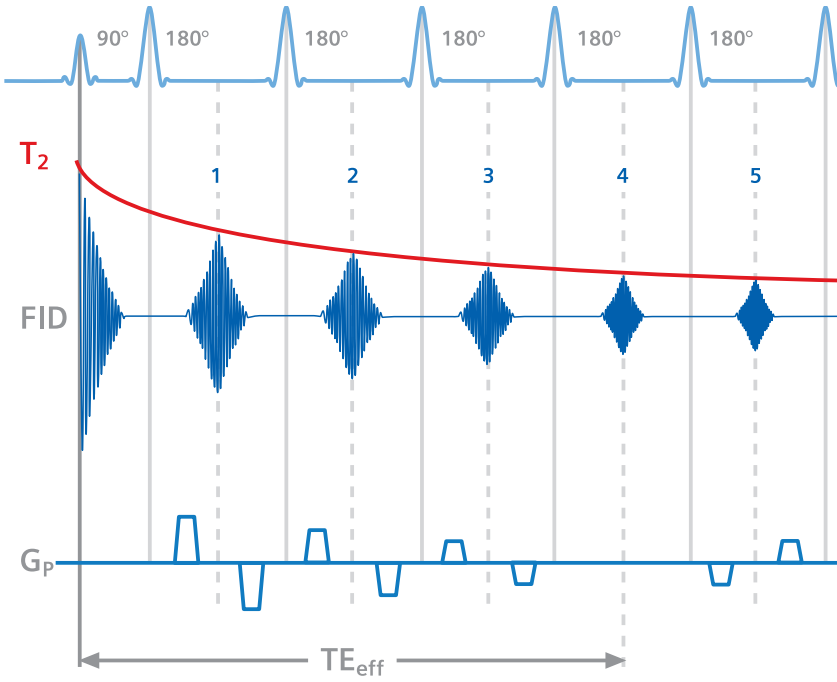
Turbo spin echo sequences (TurboSE) considerably shorten acquisition time and have replaced conventional spin echo technique to a large extent. During the time a spin echo sequence acquires not more than a single echo, the TurboSE sequence acquires an entire series of echoes.

The echo train is faster

How does a TurboSE sequence accelerate the acquisition process? With each 90 degree excitation pulse, the sequence does not generate just a single spin echo, but rather an entire series of echoes also known as the ECHO TRAIN.

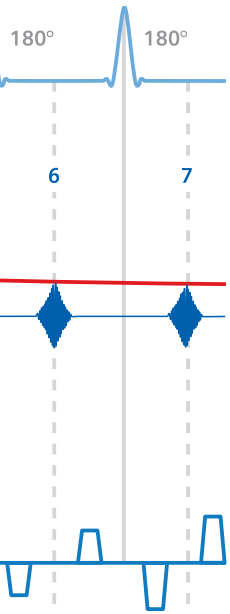
Each echo of the echo train is given a *different* phase encoding (G_P) and fills one row of the raw data matrix. The length of the echo train determines the maximum time savings or TURBO FACTOR (e.g. 7 or 15).

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 k-space

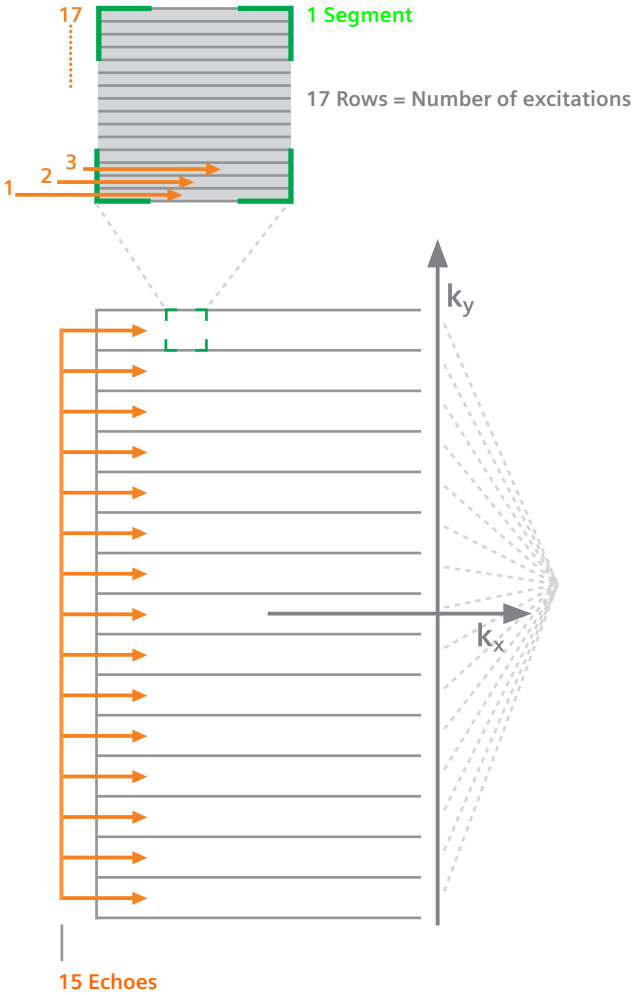
How do we fill the raw data matrix of a TurboSE sequence? For a 255 × 256 matrix only 255/15 = 17 excitations are required for an echo train of 15 echoes. This means that the sequence has to be repeated only 17 times instead of 255 times.



For this purpose, the k-space is SEGMENTED; within the repetition time TR, an entire series of raw data rows is acquired as compared to the conventional technique that requires one raw data row only.

The k-space consists of, for example, 15 segments (= Turbo factor) with 17 rows each. The total number of rows is a whole-number multiple of the echo train length (15 × 17 = 255).

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 17 times, as shown by our example.



T_2 imaging with TurboSE

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. (T_1-weighted TurboSE sequences are used, for example, for imaging the spine.)

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

To ensure detection of small hemorrhages, e.g. 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.

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

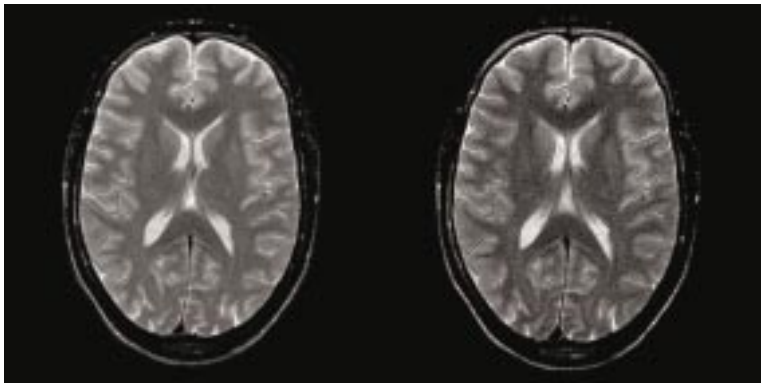


Image comparison T_2 spin echo T_2 Turbo spin echo

Review

A TurboSE sequence generates a series of spin echoes per excitation known as the echo train.

The k-space is segmented. When using an echo train of, for example, 15 echoes (= Turbo factor), only 17 excitation pulses are required. The result is a significantly reduced acquisition time.

TurboSE sequences are used primarily for T_2-weighted imaging.

Additional developments

A further development of TurboSE techniques is the combination with an inversion pulse (Turbo Inversion Recovery, TIR), a combination involving Half Fourier imaging (Half Fourier Acquired Single Shot Turbo Spin Echo, HASTE) or the addition of gradient echoes (Turbo Gradient Spin Echo, TurboGSE).

Ultra-fast with echo-planar imaging (EPI)

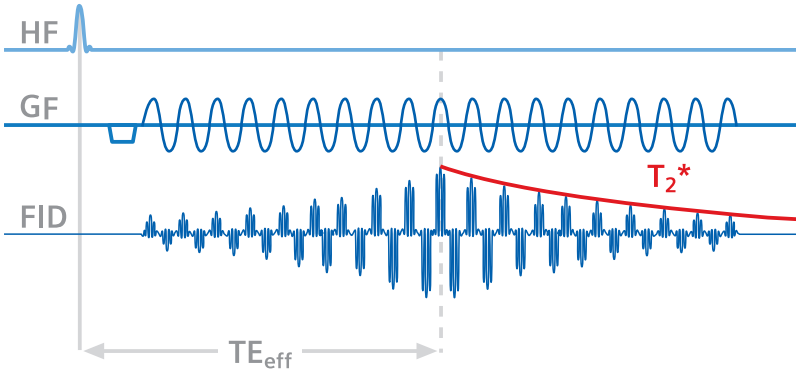
Echo-planar imaging or EPI is currently the fastest available MR imaging technique. The main characteristic of this technique is its speed. EPI acquires an entire series of images in the same time a conventional fast pulse sequence generates a single image.

It takes only a single shot ...

EPI is a SINGLE-SHOT METHOD. By that we mean that an EPI sequence uses a *single* excitation pulse to acquire an entire image.

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 128 echoes.

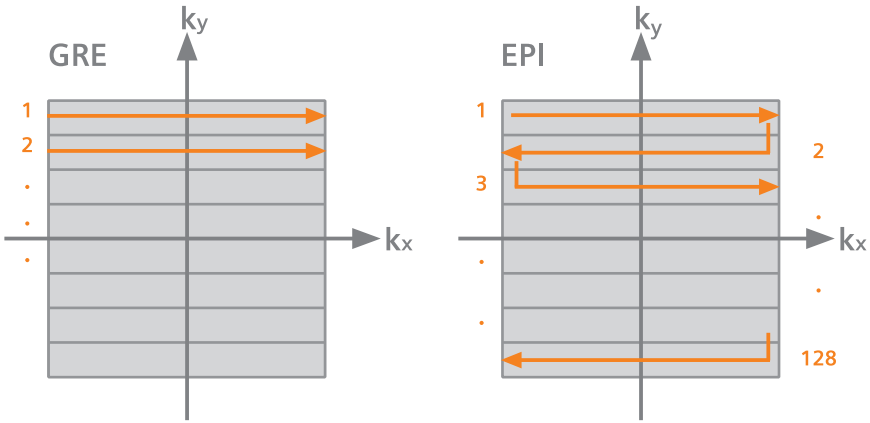


The EPI matrix is therefore between 64 × 64 and 128 × 128, accordingly the EPI factor is between 64 and 128.

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 measurement matrix is sampled in a “zig-zag” pattern (meandering).

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.



EPI is the method of choice for diffusion and cranial perfusion as well as for functional neuroimaging (BOLD Imaging).

Contrasts with Single-shot EPI

EPI is a read-out module. EPI-sequences 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.

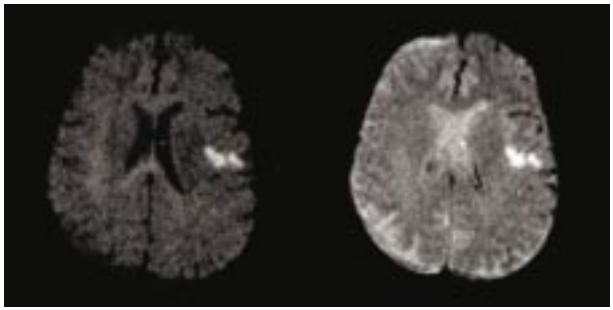


Image comparison: Strong diffusion contrast, weak diffusion contrast

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

Segmented EPI sequences

Single shot EPI sequences are very sensitive to off-resonance effects. OFF-RESONANCE means that spins outside the excited slice contribute to the MR signal and may lead to image artifacts.

These effects show up as a shift in raw data in the phase-encoding direction. This data shift increases with the echo spacing and the length of the echo train.

The echo train is shortened by sampling the raw data matrix *segment-by-segment* (as previously described in the TurboSE chapter). The shift in the phase-encoding direction is reduced and so is the visible artifact.

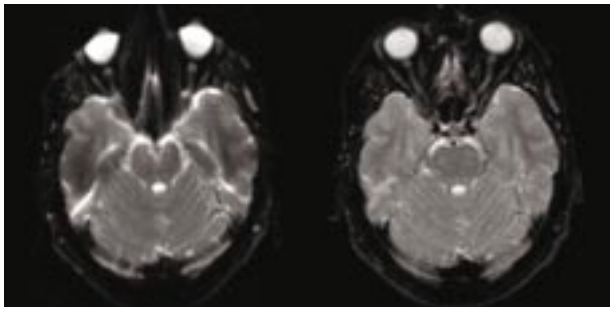


Image comparison: Single-shot EPI image with distortion artifact. The segmented EPI image (right) shows considerable less distortion in the area of the eyes.

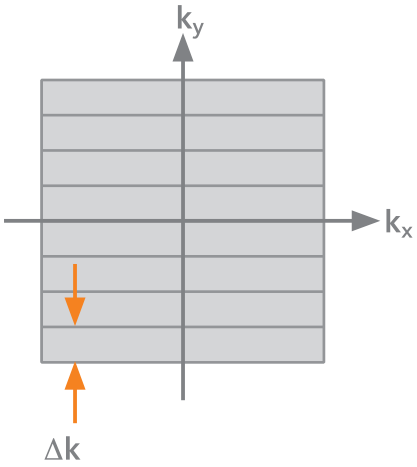
SMASH and SENSE: the parallel acquisition techniques

The speed of MR image generation is limited by phase encoding. Although the speed of pulse sequences has improved steadily with the performance of gradient hardware, the maximum switching rates of the gradients are still a limiting factor. New ways had to be found to increase speed even further. One of them is parallel data acquisition using multiple coils.

Not sequentially ...

Standard fast pulse sequences acquire data SEQUENTIALLY: they fill the k-space with raw data row-by-row (comparable to a fax machine). Each single row requires a separate application of gradient pulses. And the phase-encoding gradient constitutes a true bottleneck.

Example: to avoid motion artifacts, the patient has to hold his breath approximately 20 seconds for each exposure involving a conventional cardiac examination. This proved to be quite difficult for patients with serious heart conditions. The MR techniques introduced thus far have reached their limits.



... but parallel

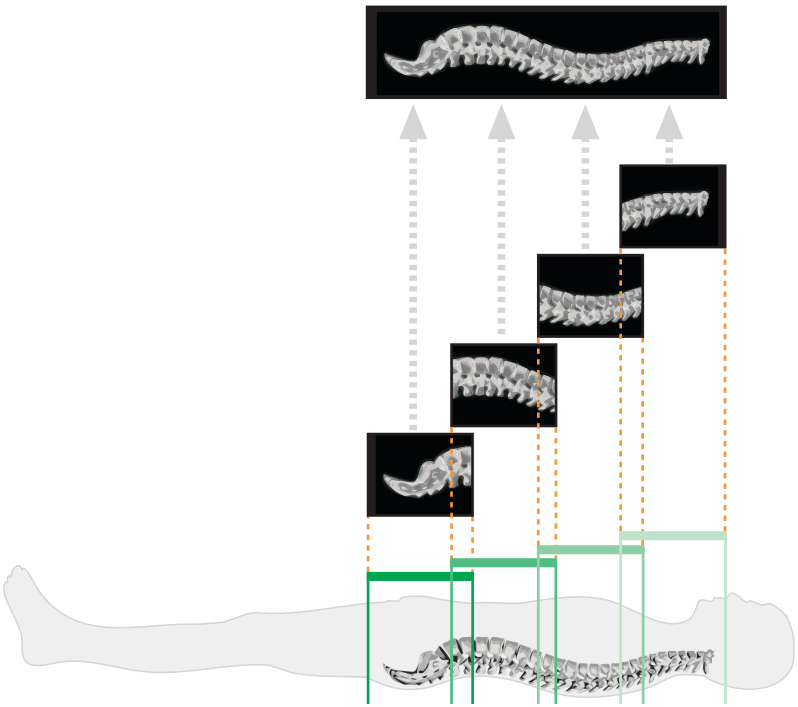
An RF coil is the receiver for MR signals. Let's assume that we would not use just one coil, but rather as many spatially arranged receivers as we would need for resolution in the phase encoding direction (somewhat similar to the methods used in a modern digital camera). In this case, we would not have to repeat a pulse sequence, but rather work without phase encoding entirely. This would considerably decrease our acquisition time ... but unfortunately, this is a futuristic goal.

Today's modern clinical techniques involve PARALLEL ACQUISITION TECHNIQUES (PAT) that use several receivers simultaneously (e.g. 4, 6, or 8). This configuration of several coil elements is known as an ARRAY. → **Arrays** have been 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 (PAT FACTOR) is 2 to 4.

The principle of array imaging

With a standard array technique, separate images are created for each coil element (in our example: 4). Theses images are subsequently combined into an overall image. For this reason, 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. As compared to standard array techniques, they use the geometric characteristics of array coils.

The spatial arrangement of the individual coil elements provides additional information with respect to 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 part of the time-consuming phase-encoding steps. In other words, we supplement the spatial encoding via the gradients with encoding via the coils.

The two most important methods are SENSE and SMASH. The difference between the two methods is that SENSE operates with image data, and SMASH operates with raw data.

The characteristics of SENSE and SMASH are slightly different. For various applications, the coils and slice orientation used determine the choice of method applied.

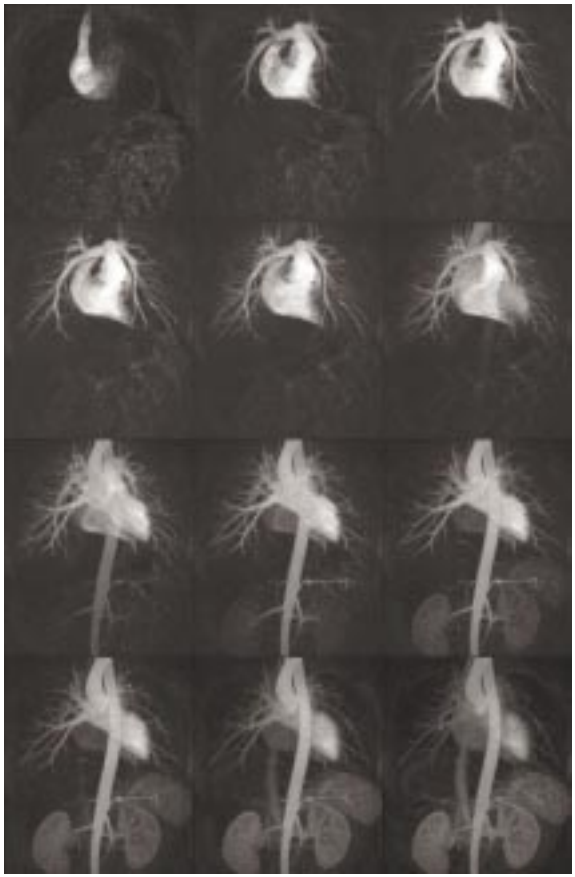
iPAT: The Siemens solution

The Siemens-specific implementation of parallel acquisition techniques is known as iPAT (integrated Parallel Acquisition Techniques).

iPAT allows either increased speed at the same image resolution or higher resolution at 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).

The echo trains of EPI sequences are shortened. The result is improved image quality, as well as less smearing and distortions in the image.



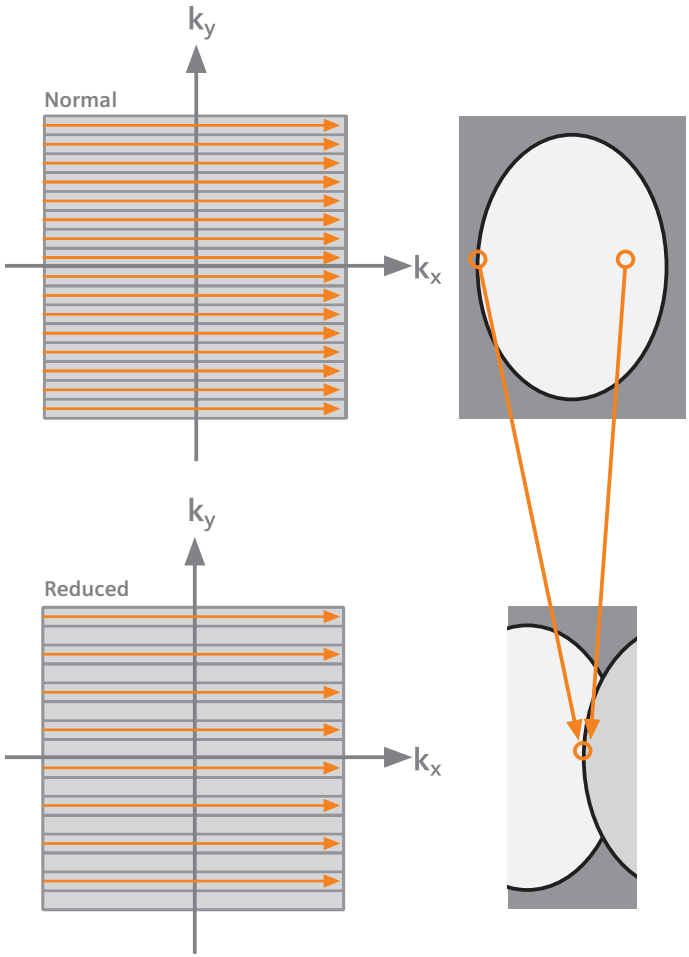
Dynamic MR angiography with iPAT. Each individual 3-D data set was measured in approximately 2 seconds (courtesy of Northwestern University in Chicago, Illinois).

SENSE: Reduce and overfold

The SENSE algorithm (*Sensitivity Encoding*) reconstructs the MR image from the image data of the individual coil elements.

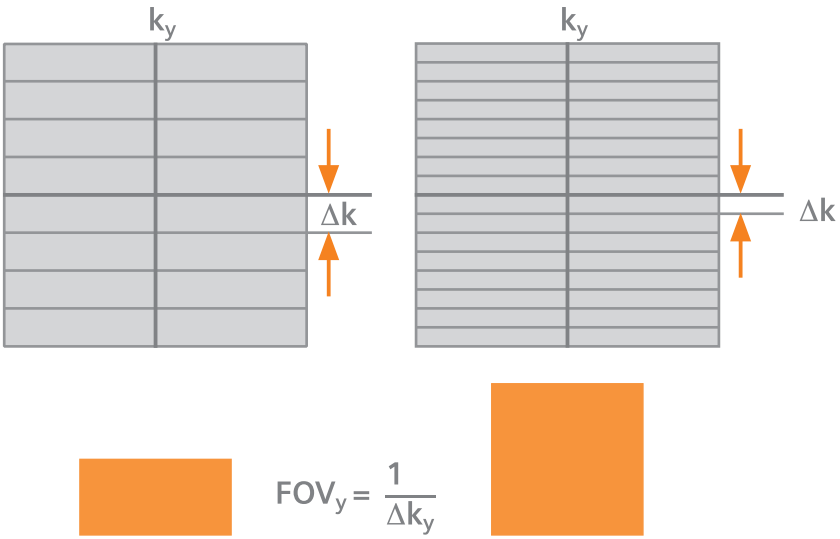
During acquisition, several phase-encoding steps are left out. For example, only every other second raw data row is filled with an echo. Basically we can look at this as an acquisition with a reduced → **field of view**. The reduced image of one coil element shows periodic overfolding in areas outside the field of fiew—similar to a transparency that has been folded up several times.

This characteristic is in principle the result of the periodicity of the Fourier technique used for frequency and phase encoding: each pixel in the reduced image is an overfolded superimposition of the pixels of the total image.



The reduced image of a phantom shows periodic overfolding in the areas outside the field of view.

Field of view, resolution, and sampling rate



The FIELD OF VIEW (FOV), is the section of the acquired slice to be shown in the image, e.g. 25 cm × 25 cm. When using a 256 × 256 matrix, each pixel has an edge length of 1 mm.
This corresponds to the maximum RESOLUTION in the image.

The SAMPLING RATE is the inverse of the field of view:

$\Delta k = 1/FOV$

In our case this means 1/25 cm. This is a phase-encoding step in units of the spatial frequency. When we make the steps larger, that is increase the sampling rate without changing the resolution, the field of view is reduced accordingly (by a factor of 2 in our graphic). However, if we keep the field of view, resolution in the image in the direction of the phase encoding will be reduced.

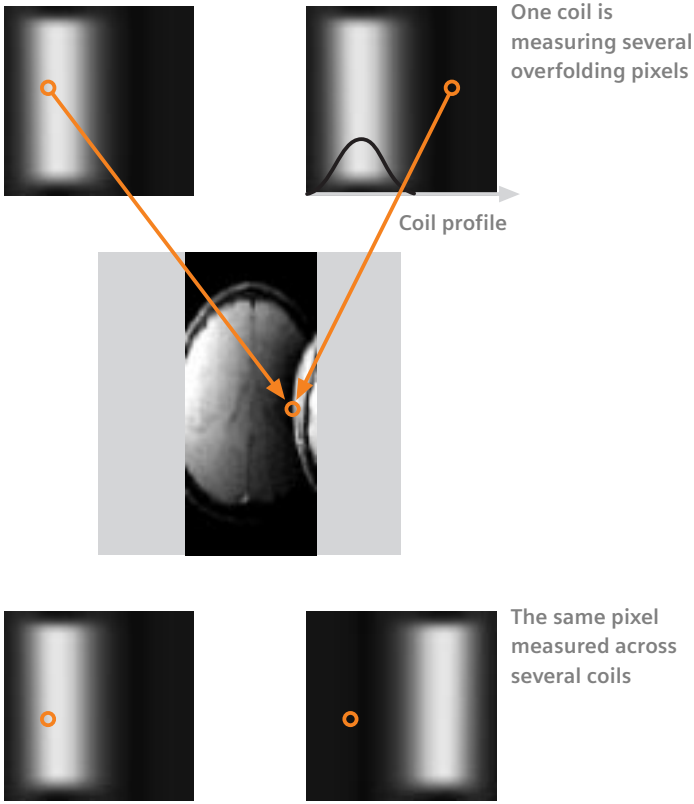
SENSE: Overfolding and unfolding

This is the difference with respect to the folded up transparency: since a coil element is *not homogeneous*, but rather has a spatial SENSITIVITY PROFILE, the overfolding pixels are not the same in the image. They are weighted with the spatial coil sensitivity.

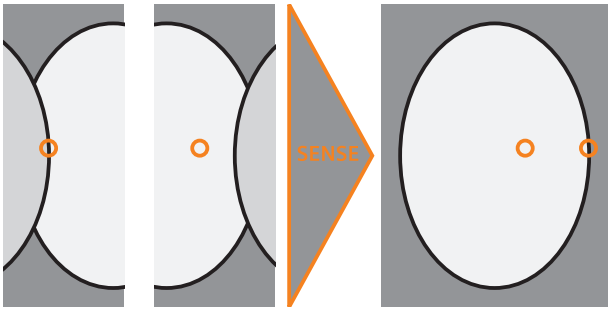
The question remains how do we locate the unfolded total image from an overfolded image.

When we have only one image with overfolding, it is not possible to unfold it properly. If possible, we can avoid overfolding through → **oversampling**. But if we acquire several overfolding images in parallel across several coils, we can reverse overfolding by using an algorithm.

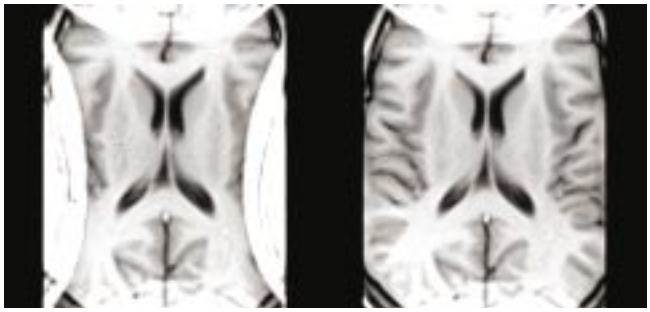
That's the basic idea of the SENSE algorithm. The implementation of the SENSE algorithm developed by Siemens is known as mSENSE (*modified SENSE*).



The SENSE algorithm computes the unfolded total image from the individual overfolding images. Pixel by pixel, the signal portions are separated from the individual localizations.

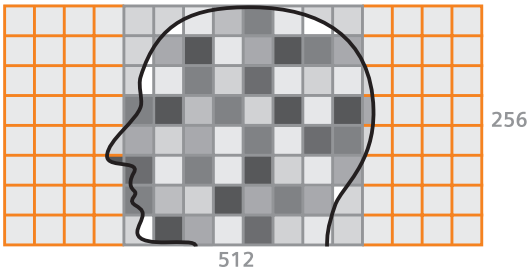


Overfolding and Oversampling



A well-known artifact: if the field of view is smaller than the size of the object, you can see the OVERFOLDING (aliasing) of structures outside the FoV.

Image comparison with respect to overfolding



By increasing the sampling rate (OVERSAMPLING), we can eliminate the overfolding effect (example: 512 instead of 256).

Principle of oversampling

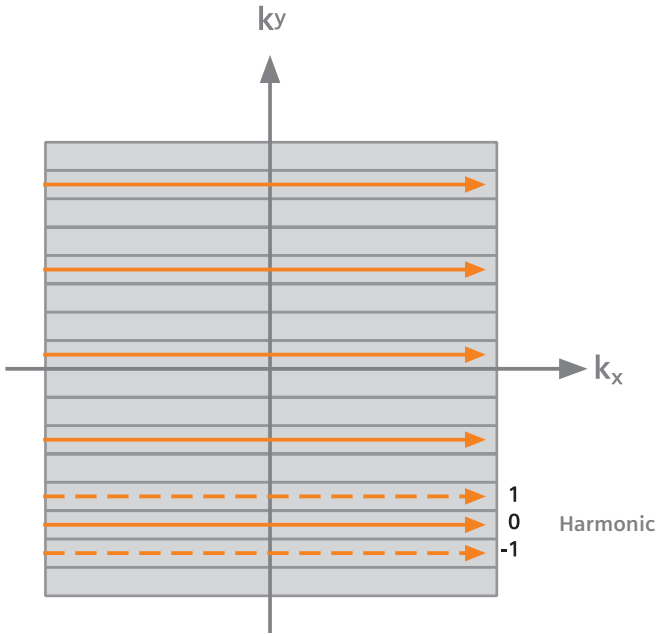
SMASH: Harmonics in the k-space

As compared to SENSE, SMASH (*Simultaneous Acquisition of Spatial Harmonics*) reconstructs the MR image from raw data.

As with SENSE, a number of phase-encoding steps are skipped. The missing raw data rows are completed by using a certain trick.

Do you still remember: the values in the k-space are spatial frequencies that correspond to stripe patterns which build up the image. The stripe patterns are nothing more than structures recurring periodically in the object to be measured. In other words: these are spatial wave patterns. The phase encoding generates these wave patterns of the spin phases.

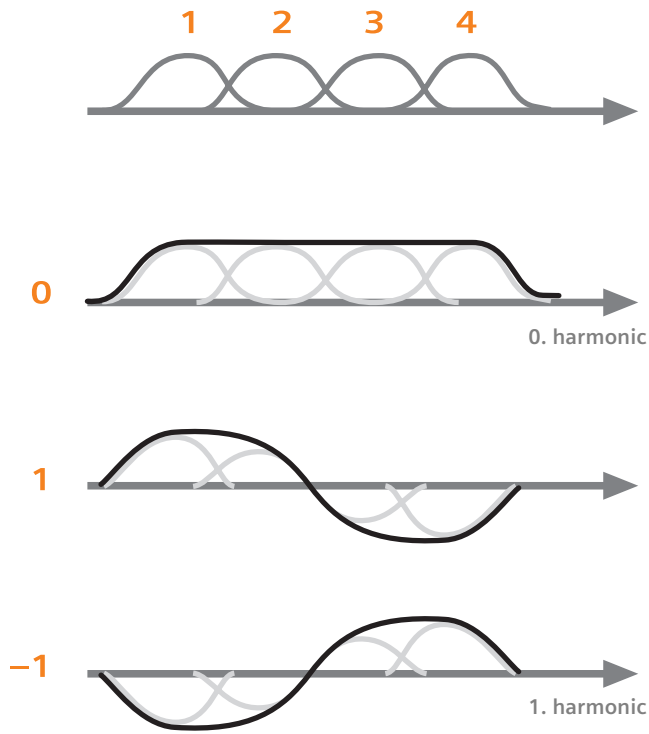
Let's assume, a receiver coil has a coil profile corresponding exactly to the wave pattern. The respective phase-encoding step would then be unnecessary. Instead of phase encoding one could, theoretically—if technically possible—increase a wavy coil profile step-by-step.



We can show the following: If a receive coil shows a coil profile in the form of a sine curve across the field of view, this sensitivity corresponds exactly to one phase-encoding step. On the basis of acoustic waves, this profile is also known as the first HARMONIC. A sine curve with double the frequency is then the second Harmonic (so-to-speak one octave higher). This corresponds to the double phase-encoding step, etc.

What's so unusual about the SMASH technique, is that we can generate the spatial harmonic through a weighted superimposition of the coil profiles of an array. With each of these harmonics, an artificial echo can be synthesized and the missing raw data is filled. We only need four harmonics to supplement the 4 missing phase-encoding steps.

The further development of the SMASH Algorithm by Siemens is called GRAPPA (*Generalized Autocalibrating Partial Parallel Acquisition*).



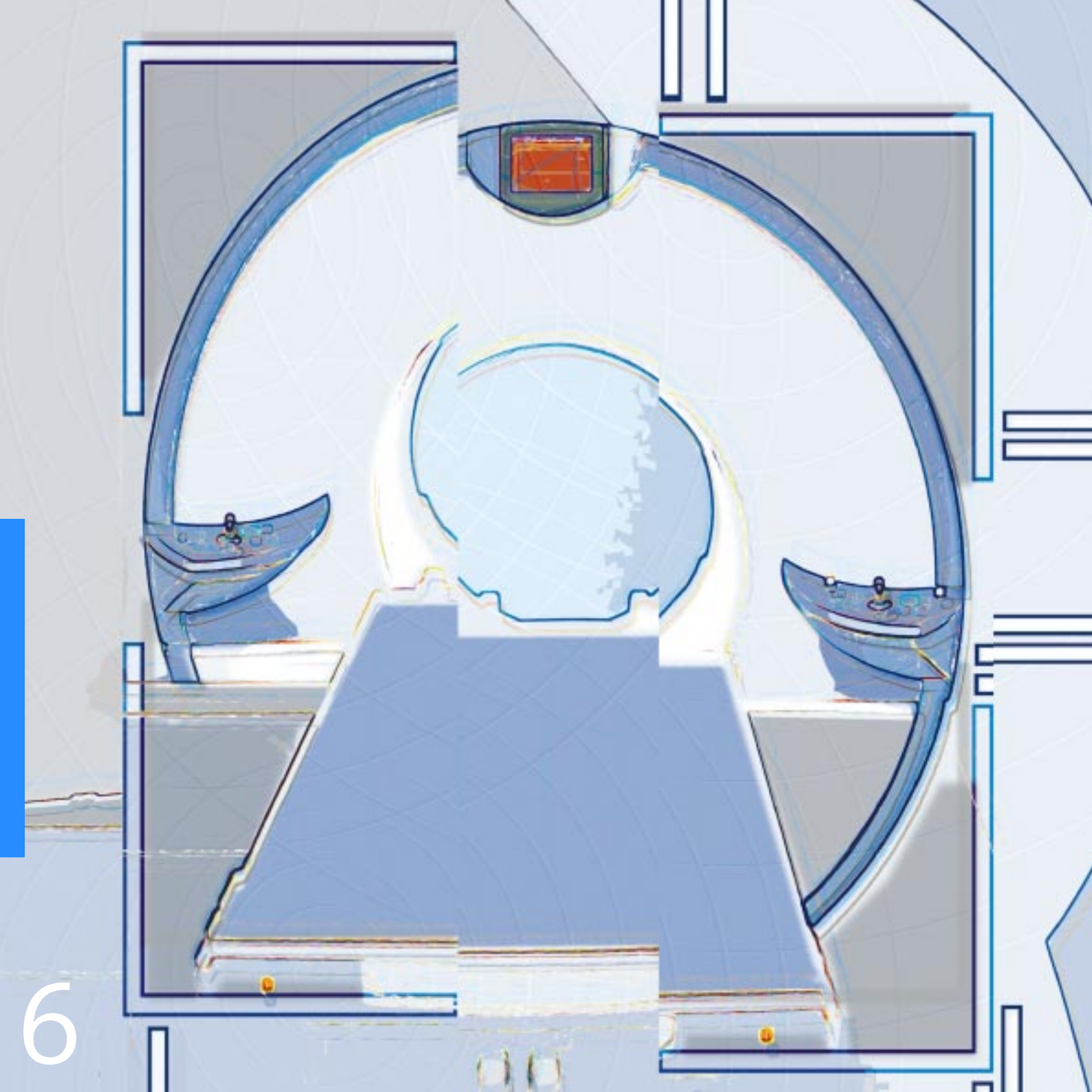
Review

With iPAT, phase-encoding steps are skipped which shortens the measurement time. In principle this is the same as generating an image of a reduced field of view with conventional imaging.

With the help of individual coil profiles the missing encoding is restored either in the image space or already in the k-space.

mSENSE
Each individual coil element generates, after the Fourier transformation of the raw data, an overfolding image. From the overfolding images, the result image is reconstructed with the mSENSE algorithm.

GRAPPA (SMASH)
At first the GRAPPA algorithm is applied to the raw data and then a completed raw data set is generated with the help of synthetic echoes. These raw data are used to reconstruct the MR image via Fourier transformation.



MR systems and their components

Today a large number of different performance-oriented MR tomographs are available for clinical applications.

The easiest way to differentiate them is by virtue of their design, since the systems do have a number of components in common.

This chapter focuses on the design of MR systems as well as their components.

System designs

Currently, the easiest way to differentiate MR systems is by their design: We differentiate between tunnel-shaped systems, open systems, as well as special systems.

Tunnel-shaped systems

Tunnel-shaped systems generate the magnetic field within what is known as the MAGNETIC BORE. These systems are known as WHOLE-BODY SYSTEMS that allow you to examine all body regions.

The advantages of this design are a highly homogeneous as well as strong magnetic field. The disadvantage of the design lies in the limited space: the patient lies in a tunnel during the examination. They may experience feelings of claustrophobia or other discomforts; children, on the other hand, may feel abandoned.

Surgical interventions during the MR examinations may be performed. However, prior to the intervention, the patient has to be moved out of the tunnel for some distance.

Open systems

The disadvantages experienced with tunnel-shaped systems lead to efforts involving open systems that are easily accessible by patients. Just as the tunnel-shaped systems, these allow for examinations of all body regions. In addition, open systems are highly suitable for interventional procedures or for, e.g., motion studies of joints.

The open MR system enables direct access from as much as three sides. Compared to tunnel-shaped systems, the limiting factors of this design are lower field strength as well as lower homogeneity.

Special systems

In the clinical field, the primary use of special systems involves examinations of extremities and joints.

Special clinical systems are characterized by low field strengths limited to their area of application.

Other special systems may be used for research (e.g. a high-field system equipped with a narrow bore for animal or test examinations).

System components

A good place to start is the measurement principle. For image generation, a homogeneous, static magnetic field is superimposed with magnetic field gradients and RF pulses. A typical MR system consists of 3 components or sub-systems: a magnet with a main magnetic field, a gradient system, and the RF system.

The magnet

The components of the magnet are located in the examination room. The room as such is shielded against interfering magnetic as well as external RF radiation. At the same time the shielding prevents the effects of magnetic and RF fields caused by the magnet on objects located outside the examination room. As a result, interferences caused by radio channels are avoided and sensitive devices are protected.

The computer system

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

- the image processor
- the host computer with console and
- the control as well as evaluation software.

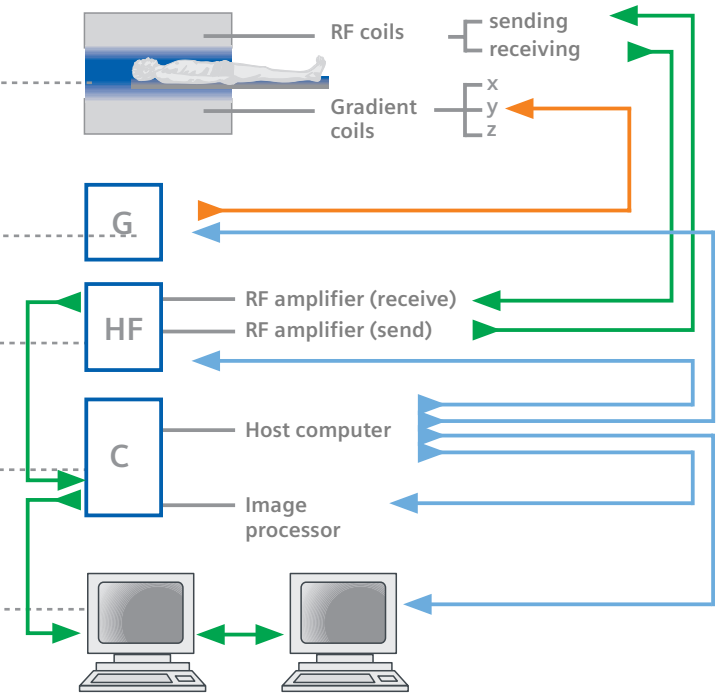
Magnet with patient table

Gradient system

RF system

Computer system

Operating and evaluation console



Documentation and data security

Both documentation and correct data are used for post-processing as well as evaluation. To secure hardcopies as well as data, they are saved for the short or long-term, depending on the tasks involved. The following may be used:

- hard disks on the host computer
- in the archiving system
- external media
  (both analog as well as digital).

The magnetic main field

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

Types of magnetic fields

Currently, the preferred two types of magnets are:

- permanent magnets with a magnetic induction ("field strength") between 0.01 and 0.35 Tesla and
- super-conducting magnets with field strengths between 0.5 and 3.0 Tesla (research systems may reach 7 T or more)

Today, normally conducting electromagnets are rarely used.

Permanent magnets

Permanent magnets consist of large blocks made from ferromagnetic alloys, e.g. in the form of a C-shaped (horseshoe) magnet.
These pole pieces are located above and below the patient. Since the material of the permanent magnet is in the pole, the main field is located perpendicularly to the long body axis. This arrangement allows for a short distance between the pole pieces and therefore a high field homogeneity. This configuration is frequently used with open systems, e.g. the MAGNETOM Concerto.

These types of magnets have a permanent magnetic field. To ensure a sufficiently high homogeneous field, they require a stable operating temperature. Since permanent magnets are not electromagnets, the operating costs are low. The obtainable field strength lies below 0.5 Tesla.

Super-conducting magnets

A super-conducting magnet is an electromagnet. A strong magnetic field is generated by the electric current flowing in large coils. The conducting wires of the coils are not made from copper as it is usually the case. Instead a niobium/titanium alloy is embedded in the copper. Liquid helium is used as the coolant, while liquid nitrogen may be used for precooling. In most cases, super-conducting magnets are applied in tunnel-shaped systems. The homogeneous magnetic field is located in the center of the magnet bore, which runs parallel to the long body axis.

Compared to an electromagnet, a super-conducting magnet is energized only *once* to the desired field strength. No additional current supply is required to keep the magnet at field.

What do we mean by super-conductivity?

At normal temperature levels, each 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.

Super-conductors are materials that have no electric resistivity at very low temperatures close to absolute zero (0 Kelvin = –273 degrees C). A constant, high current (above 400 ampere) will flow for years without an electrical potential or voltage. For this purpose, the superconductor has to be kept at very cold temperatures.

Super-conducting magnets in tube form obtain field strengths of more than 7 Tesla (high-field systems). For open systems, super-conducting magnets may be used up to a field strength of 1 Tesla.

Ultra high-field magnets

Currently, the most optimal field strength for clinical imaging is at 1.5 Tesla. In the meantime, however, tunnel-shaped systems may have a field strength of as much as 3 Tesla for clinical applications.

The advantages of ultra high-field systems are the following

- Higher image quality through an improved signal-to-noise ratio
- Shorter acquisition times effecting minimized motion artifacts as well as reduced examination times
- Further detail-enhanced exposures through higher resolutions
- Improved visualization of processes on the molecular level (Molecular Imaging)

Today, MR systems with field strengths of 7 to 8 Tesla and more represent the upper ceiling in MR. These systems are currently used for research purposes only.

Shimming the main field

The most important quality criterion for a magnet is the homogeneity of its main magnetic field. Inhomogeneities distort the spatial encoding which in turn adversely affects the slice geometry: the MR image will show distortions in the slice plane or the slice will be spatially warped.

To prevent these image errors, the magnet system has to be homogenized, that is, it needs to be shimmed in several steps. We differentiate between active or passive shimming.

PASSIVE SHIM: small iron plates are attached to the magnet for compensating inhomogeneities and distortions in the magnetic field. This means that deviations caused by manufacturing tolerances are compensated and the system is adjusted to local conditions.

ACTIVE SHIM: several SHIM COILS are attached to a shim tube in the magnet. Small static currents of different amplitude and polarity are adjusted to the shim. The small magnetic fields generated compensate for small inhomogeneities of the main field. Interferences caused in the magnetic field by patients are eliminated.

The different methods for the active shim are:

- 3-D SHIM: the shim volume is limited to the area of examination. The homogeneity will be optimized in this area only.
- INTERACTIVE SHIM: shim currents are individually set as well as optimized for the RF pulse sequence selected (used for spectroscopy).

After shimming, the main magnetic field for super-conducting magnets varies by less than 4 ppm (parts per million) within the measurement field (usually approximately 50 cm in diameter).

The gradient system

The MR system includes three gradient coil arrangements for all three spatial directions (x, y, and z). The gradient coils do not generate a permanent magnetic field. Instead they are switched on briefly during the examination.

Performance

The gradient coils are operated via special power supplies, known as GRADIENT AMPLIFIERS. These have to switch currents up to 500 ampere at great accuracy and stability. As with a loudspeaker, strong mechanical forces are exerted by the gradient coils, resulting in knocking noises during the examination. These noises are attenuated by using suitable measures.

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

Main magnetic field of magnet	Maximum amplitude (in millitesla/meter)	Rise time (in milliseconds)	Slew rate, SR (in Tesla/meter/seconds)
High-field (1–2 T)	20–40 mT/m	0.4–0.2 ms	50–200 T/m/s
Ultra high-field (3 T)	40 mT/m	0.1 ms	200–400 T/m/s

The gradient system family

The performance level of MR systems can be improved by upgrading existing gradient systems. The gradient system family known as the Maestro Class by Siemens includes four exchangeable coil systems:

- Ultra (SR: 50)
- Sprint (SR: 75)
- Quantum (SR: 125)
- Sonata (SR: 200)

To upgrade the gradient system for an existing MR system is a cost-effective solution to meet increasing requirements. This upgrade does not affect the magnet which is the most expensive component of the system.

The radio-frequency system

The nuclear spins of the body tissue are stimulated by pulsed magnetic RF fields. These RF pulses are *transmitted*, that is, the MR signal received from the spins has to be *received*. The RF system meets this requirement.

Overview

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

- RF antennas (coils)
- RF transmit amplifier
- RF receive amplifier

RF antennas (coils)

The transmit and receive antennas used for stimulating resonance are known as COILS or *resonators*. These coils come in all shapes and sizes.

The BODY COIL is an integral part of the MR system. Its function is that of a whole-body antenna system. Additionally, the body coil has a large measurement field.

Special coils

Depending on the body region to be examined, additional SPECIAL COILS are connected and positioned locally on the patient's body. The shape of the special coil is determined by its area of application.

To increase the effect of a coil, the FILL FACTOR, the ratio between the sensitive volume of the coil and that filled by the patient, should be as large as possible.

Homogeneity of the RF field

For a transmit coil, the homogeneity of the RF field in the stimulated volume is an important quality criterion. The level of stimulation applied to the atomic nuclei involved should be same. For example, saddle or cylinder-shaped coils are best for the homogeneity of the RF field for the magnetic main field of magnets with an axial horizontal field axis.

Signal to noise

The RF receive coil not only receives the desired MR signal. It also receives unavoidable NOISE. The primary source for the noise is the Brownian molecular motion in the measured probe, in this case the patient. The noise depends largely on the size of the coil. In short, the larger the coil, the greater the noise. For this reason, small local coils have a better signal-to-noise ratio if used with a correspondingly smaller measurement field.

LP and CP coils

RF waves are usually POLARIZED, that is, they oscillate in one plane. Depending on the polarization, we differentiate between

- LP coils (linearly polarized) or
- CP coils (circularly polarized)

A CP coil has a better signal-to-noise ratio than an LP coil.

Array coils, Integrated Panoramic Array (IPA)

Array coils are used when examining larger measurement areas. They combine a high signal-to-noise ratio with the measurement fields of large coils. To this end, array coils include several independent smaller coil elements that can be combined according to the area under examination.

Siemens developed this system further into an *Integrated Panoramic Array (IPA)*. Depending on the system, IPA allows you to simultaneously switch up to 16 independent CP coil elements. This enables you to examine different regions of the body (e.g. head, neck and spine) in a single measurement without time-consuming coil exchanges. As a result the length of examinations per patient has been considerably reduced.

The RF transmit amplifier

The RF transmitter has to meet the most stringent requirements: during the entire measurement, the transmitter has to accurately send RF pulse sequences of varying center frequency and bandwidth. For this reason, amplification includes two stages:

- the preamplifier generates the signal
- the transmit amplifier increases the signal gain as required

The RF receive amplifier

After it has been received, the very weak MR signal is amplified in an extremely low-noise amplifier before it is digitized and processed further. The better the signal, the stronger and clearer it will be received by the coil. The signal strength depends for one on the volume excited in the receive coil, and for the other on the distance to the measurement object.

The computer system

In the past, the computer system was located in a separate room. Today, the individual parts are highly compact and can be stored in containers located under the table in the control room.

The image processor

The amplified MR signal has to be digitized prior to being processed further for image reconstruction.
An ANALOG DIGITAL CONVERTER computes the analog signal into digital, finely scaled individual values: it is sampled at fixed intervals (in µs). The computer system is now able to process the digitized measurement values.

MR image reconstructions using a 2-dimensional Fourier transformation is a time-consuming process. A state-of-the art image processor is able to reconstruct approximately 100 images/per second using a 256 × 256 matrix.

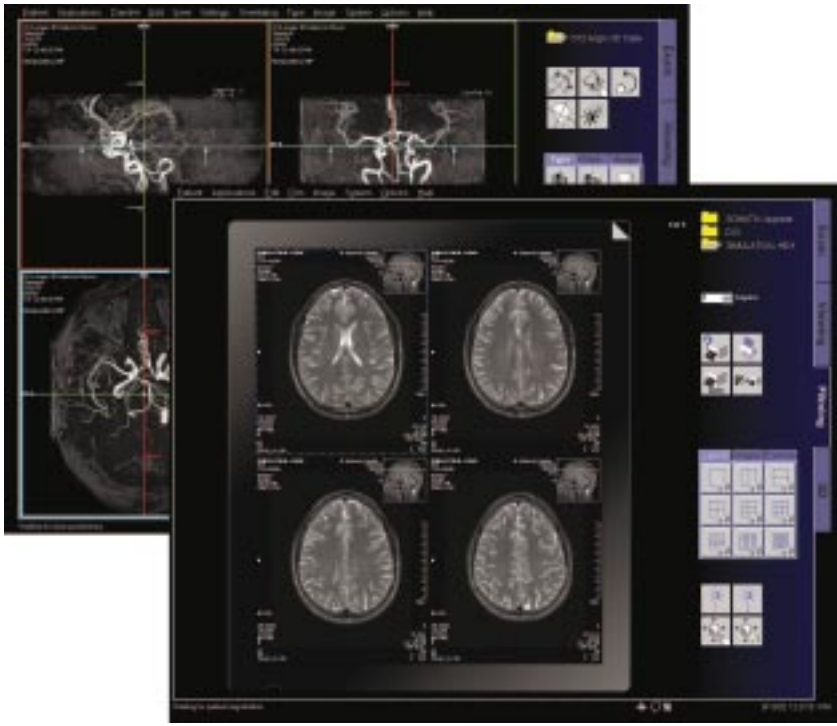
To be able to take full advantage of the processor's output range, all of its other components have to be dimensioned accordingly. The main memory (RAM) is usually in the Gigabyte range. The raw image data is stored on fast, high-capacity hard disks.

The host computer

The host computer controls and monitors the entire system (data input, measurement sequence, image display).

The host computer is a high-performance multipurpose computer that includes several state-of-the art processors for quickly processing either different or parallel tasks. The type of host computer used determines the response time of the system to user commands as well as data entries.

Since a high-performance host computer allows you to simultaneously process several tasks (multi-tasking), you are able to evaluate and process the first results on-screen while the examination is still in progress.



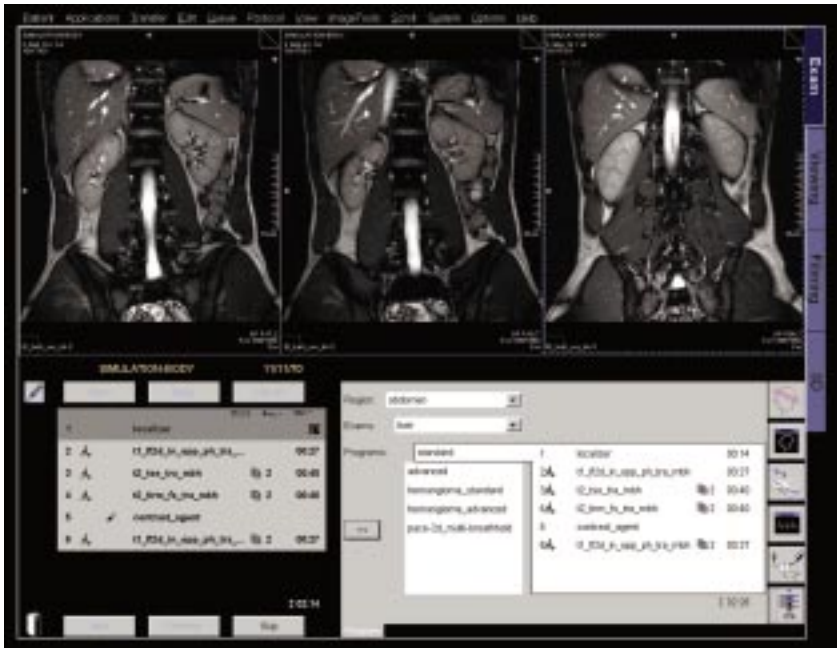
Control and evaluation software

The system requires a high-performance control and evaluation software that acts as the interface between the MR system and you. The software is of modular design and includes the following:

- patient management
- organization and control of measurement system
- measurement data acquisition and processing
- display of image data
- image post-processing
- image documentation and data security

The high user-friendliness obtained is based on ergonomic user interfaces as well as easy operation.

The control software handles the organization and control of the measurement system. This includes automatic functions such as 3-D shim as well as integrated measurement programs. As the user, you are able to select measurement protocols optimally adjusted to the examination or you can modify and store them for future use.



The evaluation software allows you to process and evaluate new images while the examination is still in progress. You are also able to select stored images at any time and post-process them.

Image processing includes:

- windowing: you select window width and position, as well as automatic contrast optimization, etc.
- automatic Cine run: allows you to rapidly page through images
- statistical evaluation: surface area determination, measuring distances and angles,
- 2-D post-processing: mirroring, image annotation, image zoom as well as shift, etc.
- 3-D post-processing: views in any direction, 3-D display of surface areas, etc.
- dynamic evaluation: addition of images, evaluation of contrast agent studies, computation of T_1/T_2 images, etc.

MR users may utilize new evaluation functions by updating the software or buying the respective option.

Integrated post-processing (inline)

Use of inline technology allows for real time processing during image reconstruction

- image subtraction
- MIP (Minimum Intensity Projection)
- standard deviations
- storing of original images
- diffusion imaging, computation of: trace weighted images, ADC maps (Apparent Diffusion Coefficient), Global Bolus Plot (GBP), Time-to-Peak (TTP)
- BOLD imaging (Blood Oxygen Level Dependent): z-score (t-test) computation, spatial filter, ART (Advanced Retrospective Technique) for fully automatic retrospective motion correction.

Special evaluations

Modern evaluation software offers numerous possibilities for special evaluations:

- MPR (Multiplanar Reconstruction)
- MIP
- MR spectroscopy including metabolite images and spectral overview cards
- evaluation of time dependencies (MTT = Mean Transit Time, Mean Curve)
- SSD (Surface Shaded Display)
- 3-D VRT (Volume Rendering Technique)
- Image fusion
- Vessel View
- BOLD evaluation
- Neuro perfusion evaluation (TTP, relMTT, etc.)
- Image filter
- Argus: evaluation of cardiac functions, flow quantification, evaluation of time dependencies

Image documentation and data security

The advances made in digital technology enable fast and extensive data storage. Considering MR image processing and its large data volume and archiving, MR imaging has profited greatly from this new technology.

The hard disk

The evaluation processor includes a separate hard disk for storing image data. The disk is used as intermediate image storage during processing. During the last year, the capacity of hard disks has reached a level that allows us to store more than 100,000 images in a 256×256 matrix. As a result, images designated for post-processing can be stored for several days since they are not affecting the storage capacity available for new examinations.
We do recommend, however, to use the hard disk of the evaluation processor for intermediate storage only.

The archiving system

A long-term storage system is used for archiving images. They are either burned on CD's or DVD's in the future. A CD can hold up to 4,000 images in a 256×256 matrix. A DVD may store up to 52,000 images of the same matrix size.

A jukebox allows for easy data management of images archived on a CD or DVD. As an external system, the jukebox is connected to the computer system and includes a storage as well as a playback system. Additionally, CD's are burned in the jukebox. The storage capacity of a jukebox is as high as 150 CD's or 255 DVD's. For data volumes of this size, a software module with quick access to stored image data handles the requirements of automatic registration and identification.

External media

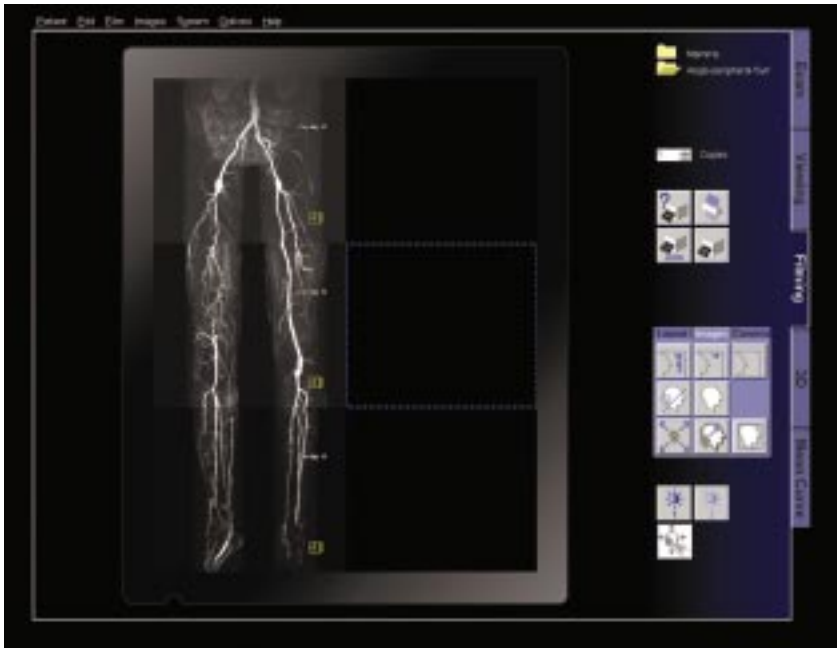
In some instances, it may be necessary to store image data to external media, for example, on a single CD. However, this approach is frequently replaced by increased networking via the Internet or medical networks.

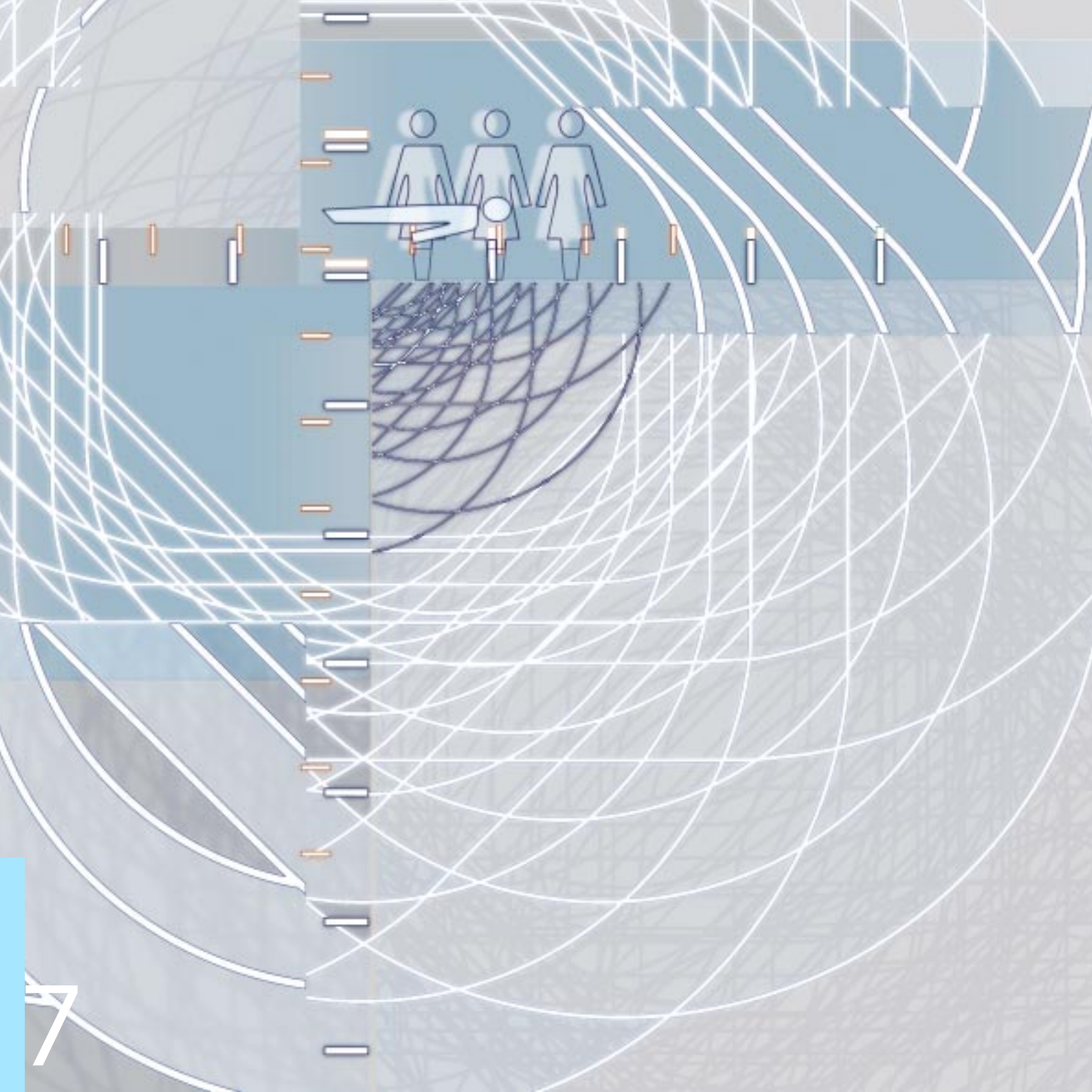
Since a computer system may not be available to evaluate these digital image repositories, the MR images may be displayed on analog media:

- X-ray film: a laser camera connected to the system copies the images with high resolution to an X-ray film
- Paper: paper copies are made using a laser printer. They have a lower resolution than X-ray films and are not especially suited for evaluation.

DICOM

DICOM (Digital Imaging and Communication in Medicine) is a standard for electronic data transfer of medical images. This standard allows for communication between devices by different manufacturers.





Environmental as well as biological effects

To the best knowledge of the technical as well as scientific community, MR examinations do not present risks to patients when the MR system is operated as specified. Examinations may also be repeated without risk.

The safety requirements that need to be observed when operating the MR system or examining the patient are listed on the following pages. The areas designated as hazardous are: the static magnetic field, the magnetic fields changing over time (gradient fields) and the RF field applied.

Static magnetic fields

The strong static magnetic field of a Magnetic Resonance Tomograph is applied to align the nuclear spins of the tissue to be examined. The strong field affects the tissue as well as all other magnetizable material in the vicinity of the magnet.

Biological effects

Since the introduction of MR tomography, a number of examinations have been performed to determine the biological effects of the static magnetic field. The effects known include, for example, dizziness, stomach-upsets as well as a metallic taste. Most of these effects occur only after field strengths above 3 Tesla. These are *short-time effects*, that is, they occur exclusively in the magnetic field or shortly after leaving it. To date, biological long-term effects have not been observed.

According to state-of-the art knowledge, examinations with static magnetic fields up to 4 T are without long-term effects.

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

Magnetic effects on devices and material

Magnetizable materials, e.g. such as iron, are attracted by the magnetic field of 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 accelerating them when moving them 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 pace makers or hearing instruments. As determined by national and international recommendations and guidelines, the safety/exclusion zone for pace makers has been established at a field strength of 0.5 m T.

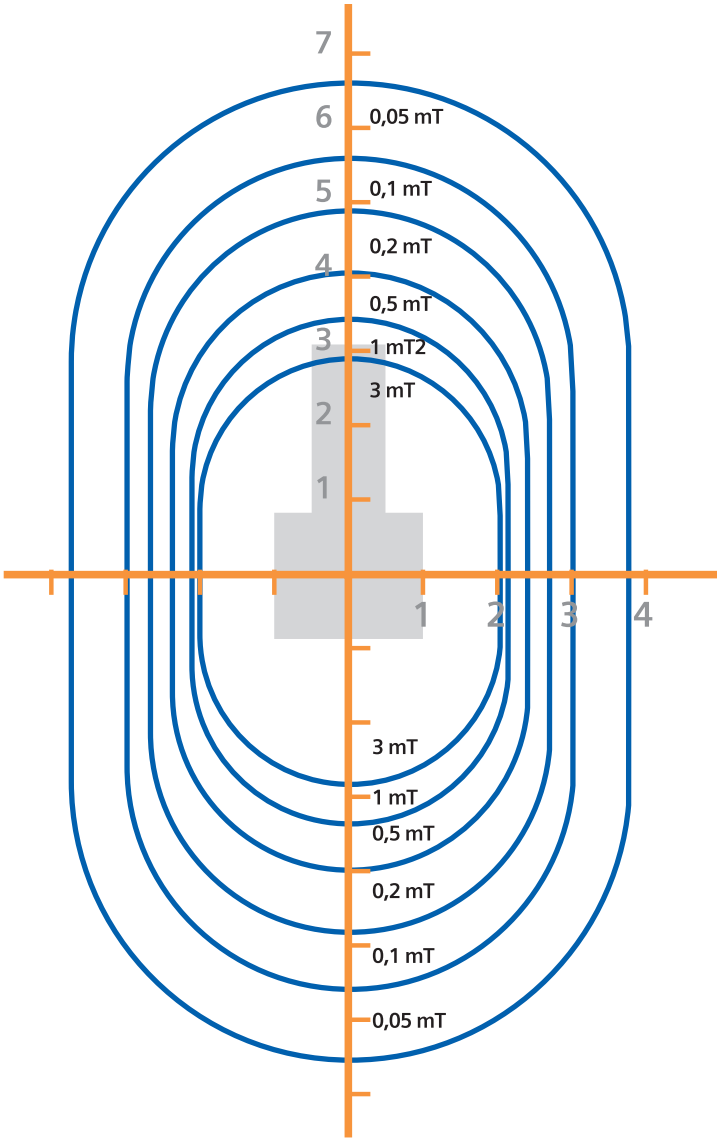
The functionality of hearing instruments may be comprised in strong magnetic fields.

In each case, the patient has to be questioned 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 1.5 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

One of the advantages of permanent magnets is the low fringe field, since the system is usually designed with a flux return and is also largely self-shielding.

For super-conducting magnets, the fringe field is shielded by using additional 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 opposite direction on the field generating coils.

Time-varying magnetic fields (gradients)

In addition to the static magnetic field, time-varying gradient fields are applied during MR examinations. These generate electrical voltages as well as currents (law of induction) in conducting materials, in this case in the human body. These currents are very small and usually do not present a source of hazards, for e.g. the heart.

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 contracted involuntarily are not hazardous to the patient's health, but may however, be uncomfortable for the patient.

In the safety standard for MR systems IEC EN 60601-2-33, the maximum field changes are defined as a function of the switching time. They reach a magnitude of 40 T/s during fast sequences (for a switching time of e.g. 400 µs).

Normally these threshold values are not exceeded by the imaging methods used today. The only time the stimulation effect may be exceeded is with extremely fast gradient switching during EPI. For safety purposes the gradient pulse has been limited.

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.

Noise

Noise is generated in the MR system when switching the gradient fields at high speed. Depending on the type of system or measurement sequence, ear plugs should be worn.

Radio-frequency fields

The electro-magnetic RF fields used in magnetic resonance lie in the frequency range of radio waves. Three different safety aspects should be observed: tissue warming, interference superimposed on other devices as well as external interferences.

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) as well as 8 W/kg (spatial peak).

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 (e.g., 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 interferences (e.g. radio, cell phone, 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).

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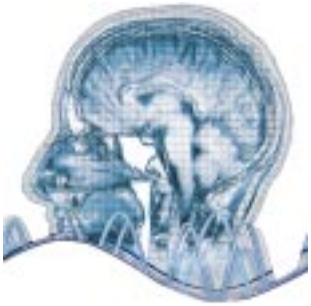

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