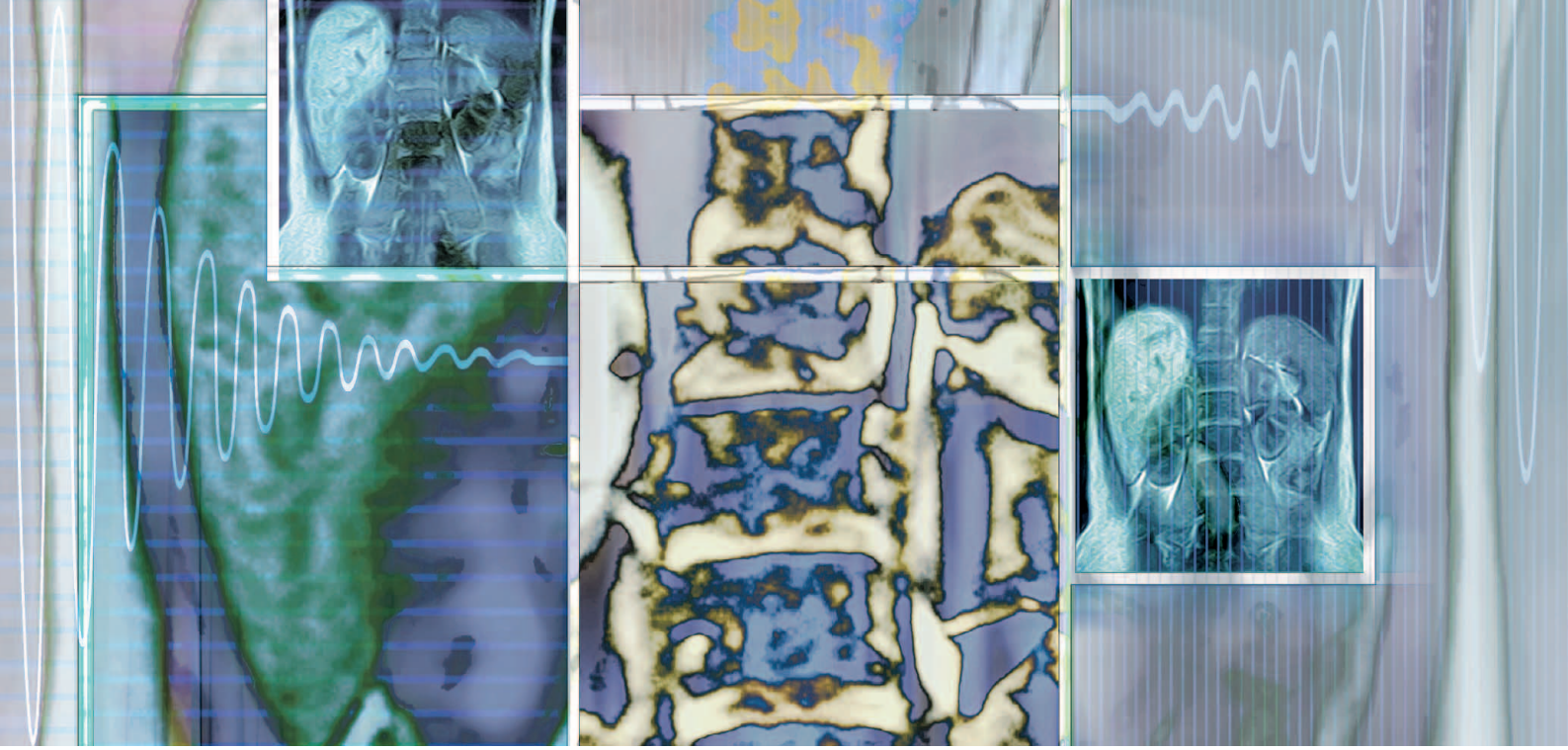

Magnets, Flows, and Artifacts

Basics, Techniques, and Applications
of Magnetic Resonance Tomography

SIEMENS
medical

Magnets, Flows and Artifacts

Magnets, Flows and Artifacts

Basics, Techniques, and Applications
of Magnetic Resonance Tomography

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Magnets, Flows and Artifacts

This is the second volume of *Magnets, Spins and Resonances*. This volume continues with the basics of magnetic resonance tomography. Until now we have limited ourselves to the excitation of stationary spins. Now we are going to let the spins move: We will be talking about flow effects, saturation, functional imaging, and artifacts.

The book ends with an introduction to image quality and the descripiton of additional fast imaging techniques.

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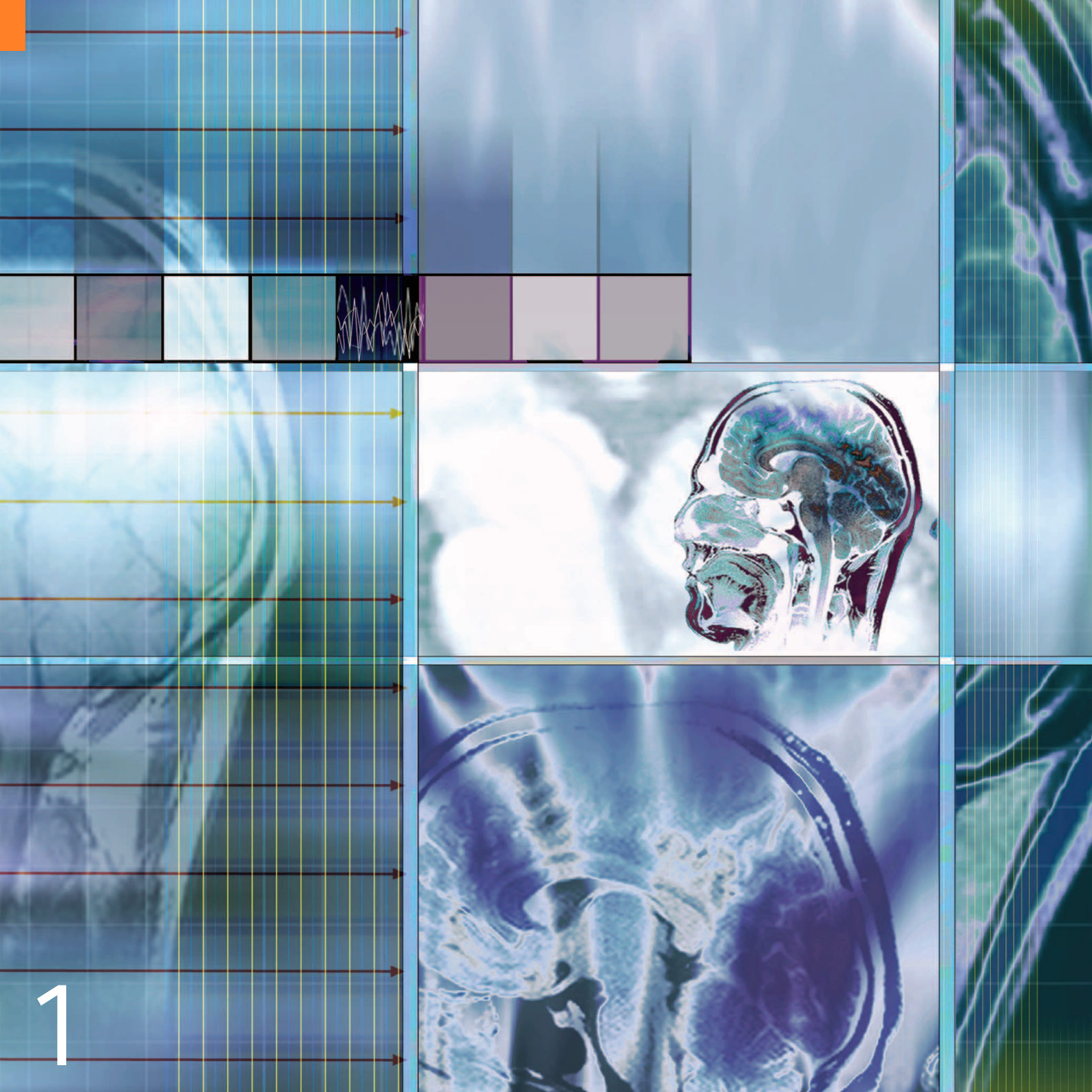
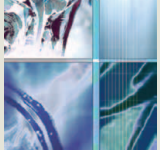


1

Image quality

Image quality is the beginning and end of MR imaging. It is the product of complex interactions between the spatial resolution of the structures acquired in the image and signal strength and contrasts obtained in relationship to unavoidable noise.

The secret lies in optimizing image quality in relationship to the measurement time required.



Signals, noise, and contrast

One of the objectives of the MR examination is to obtain the diagnostic image information necessary in the shortest possible measurement time. The image quality has to be of diagnostic usefulness. How can we affect image quality? What constitutes a good image?
The most important criteria for image quality are: A strong signal, low noise, good contrast as well as sufficient resolution.

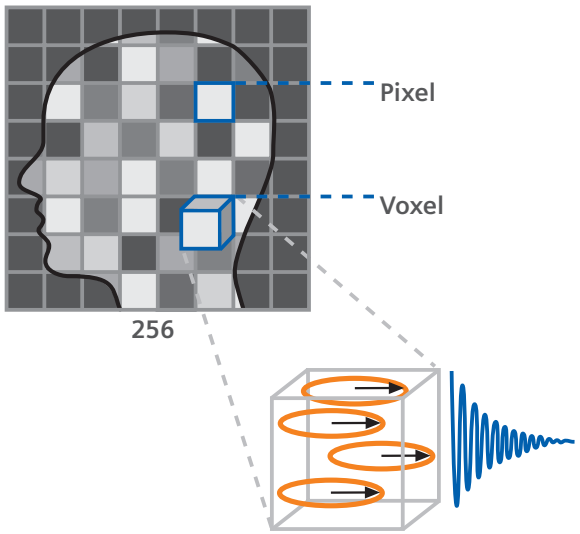
How do you generate an image from a signal?

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

After the RF pulse stimulates a slice in the patient's body, each voxel in this slice emits an MR signal.

Among other things, the SIGNAL STRENGTH depends on the quantity of signal-generating proton spins in the respective voxel (proton density).

> The more spins contribute to magnetization, the stronger the signal.



This is how contrast is generated

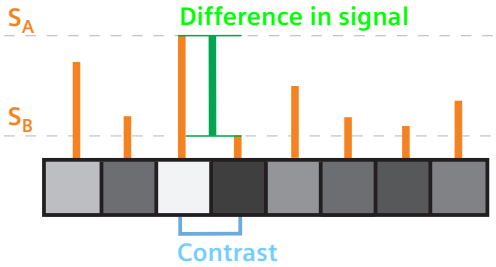
The raw data signal acquired from the slice is a mixture of these individual voxel signals. The MR image is computed from the raw data signals using Fourier Transformation. In this manner, each voxel can be assigned a signal intensity and therefore a respective grey value.

Bright pixels in the image represent stronger signals, weaker signals result in darker pixels. For simplicity's sake, our example shows a single row of 8 pixels.

The CONTRAST in the image is, to put it simply, the difference in signal strengths between two types of tissue, A and B. In other words, contrast is equal to the difference in signal:

Contrast = Difference in signal = $S_A - S_B$

Each type of tissue emits individual signal strengths. This allows for anatomical differentiation in the image and, in the final analysis, differentiation between pathalogical and healthy tissue.



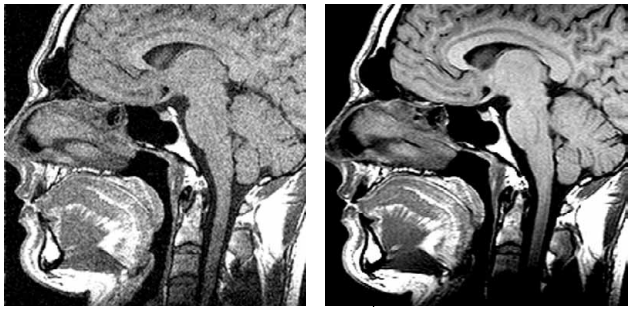
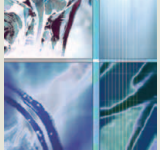


Image comparison: Low T_1-contrast (left), high T_1-contrast (right).



Signal versus noise

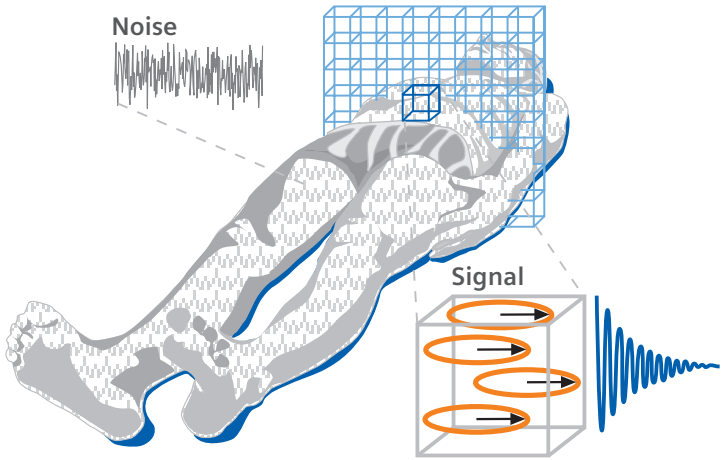
Noise is an unwanted occurrence in the MR image. Noise is basically superimposed on the MR signal. Noise appears in the image as a grainy, random pattern similar to snow on the TV screen. The only difference is that it is rigid and fixed. This phenomenon deserves our attention because of its strong adverse affect on image quality.

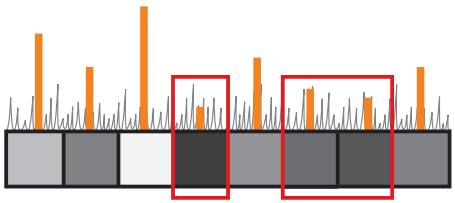
When the signal is drowned in noise ...

The NOISE in the image represents statistical fluctuations in signal intensity that do not contribute to image information. What is the source for this effect?

As discussed previously, the MR signal is emitted by the selected slice and/or the respective voxel. In comparison, noise is generated throughout the human body through the molecular motion of charged particles. This is known as the Brownian motion. To that we add the electronic noise of the receiver technology.

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





... the relationship is off

An important criterion for MR is the SIGNAL-TO-NOISE RATIO (SNR):

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

A higher SNR means improved image quality.

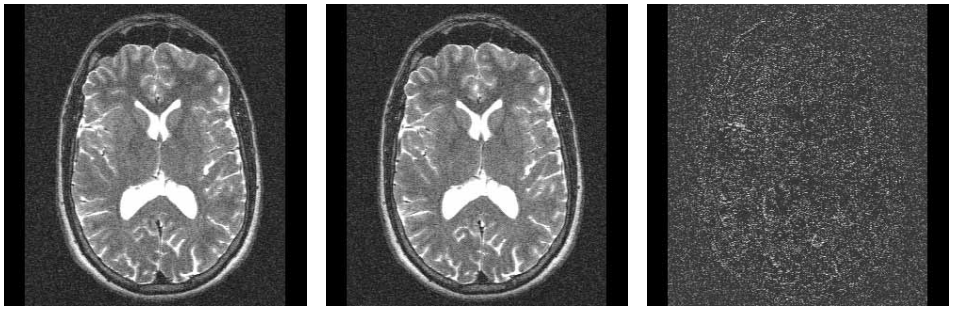
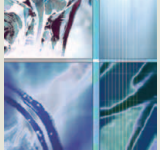


Image comparison: The two images to the left were acquired the same way and subsequently subtracted from one another (subtraction = pixel-by-pixel difference in grey values). What remains is background noise (to the right).



Amplifying the signal

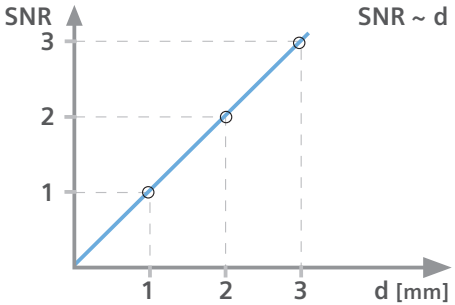
It is not possible to suppress the noise in the image. However, the signal can be amplified. A strong signal is therefore the first step toward good image quality.

... through thicker slices

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

The SNR is directly proportional to the voxel size.

Disadvantages: Increasing slice thickness reduces spatial resolution. And the results may be partial volume effects that distort image results (e.g., bones protruding into soft tissue).



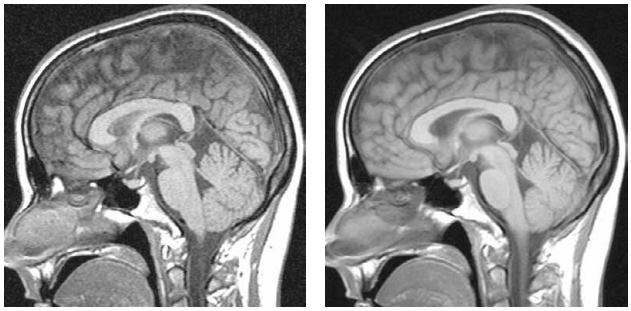


Image comparison: The right figure shows a slice that is three times as thick as the one in the left image. Results: The SNR is three times what it was.

... through more acquisitions

We do not have to select an undue large slice thickness. The SNR can be improved through other methods as well: through multiple measurements of one slice (several ACQUISITIONS) and through averaging the results in a single image.

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

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

For example: When measuring and averaging 4 acquisitions in slice, total SNR is now twice what is was before.

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

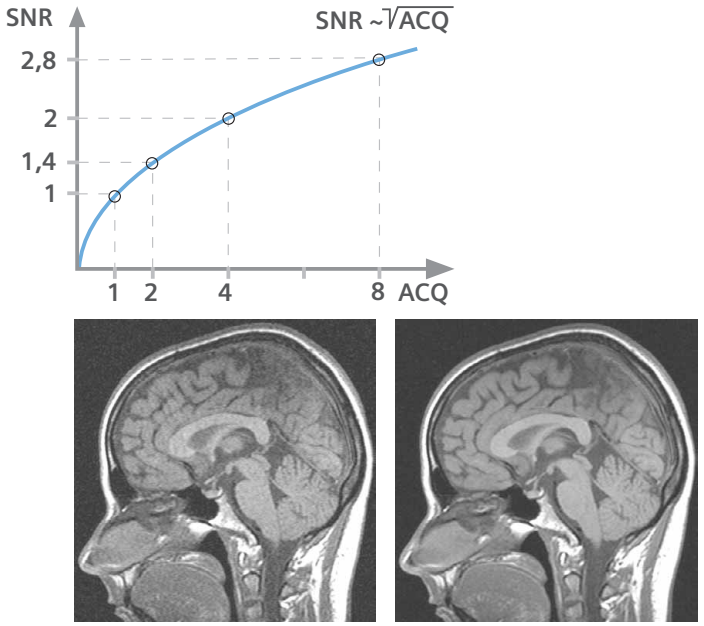
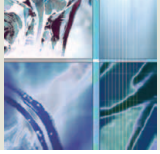


Image comparison: left 1 measurement, right 4 measurements. Results: The SNR to the right is double the SNR to the left.



This is how contrast and noise are connected

A high signal-to-noise ratio alone does not guarantee easy differentiation of two structures in the image. They also have to show sufficient contrast. This leads us to a combined as well as important quality criterion, the contrast-to-noise ratio.

What does contrast have to do with noise?

As a reminder: Up to now we have defined contrast quite simply as the difference in signal between two tissue types. This difference is visualized by the different pixel grey values. In reality, however, the effective contrast is always related to the noise level.

The CONTRAST-TO-NOISE RATIO (CNR) in the MR image is the difference between the signal-to-noise ratios of two relevant tissue types A and B:

$CNR = SNR_A - SNR_B$

This is the contrast we actually see and evaluate in the image.

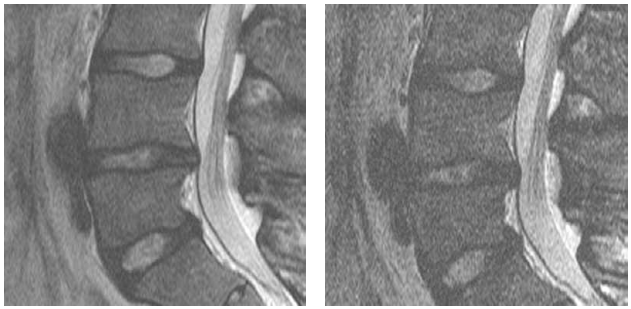


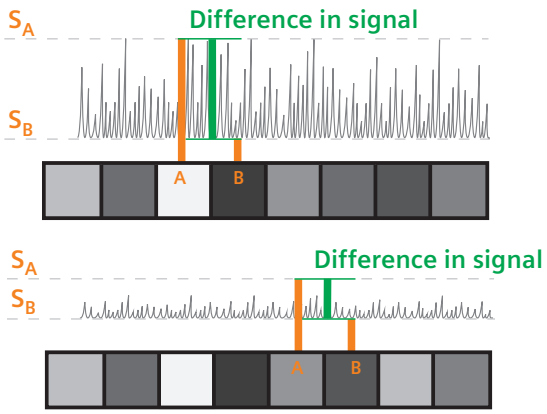
Image comparison: good CNR, poor CNR.

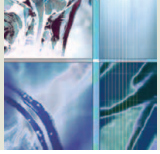
When the contrast is too noisy …

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

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

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





Reduced to the essentials

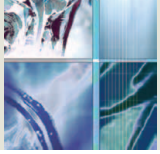
SNR and CNR are important criteria for MR image quality. They create a relationship between the signals or contrasts in the image and the portions of noise.

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

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

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

The contrast-to-noise ratio (CNR) is the difference of the signal-to-noise ratios between two relevant tissue types. Since CNR equals effective contrast, it is a better quality criterion than SNR.



Measurement matrix and resolution

Contrast and signal-to-noise ratio determine image quality. They are a basic but not sufficient for an accurate diagnosis. Let's take a look at the spatial resolution of the structures acquired in the image.

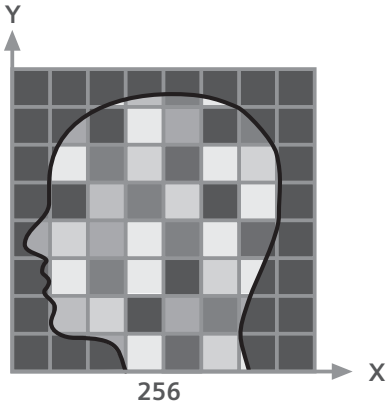
The matrix determines resolution

With a square measurement matrix, the number of columns (BASIC RESOLUTION) is the same as the number of rows (PHASE RESOLUTION).

The phase resolution determines the measurement time: When you cut phase resolution in half, you get half the measurement time as well since the number of time-consuming phase-encoding steps is also reduced by 50%. When using twice the phase resolution, the measurement time increases accordingly by 50%.

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

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



Matrix size and signal-to-noise ratio

The matrix size not only determines resolution, it also affects the signal-to-noise ratio.

Let's remember: The larger the voxel selected, the stronger the signal. Because the more protons you have that contribute to magnetization.

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

Matrix	Relative SNR
128	1.4 ($\sqrt{2}$)
256	1.0
512	0.7 ($1/\sqrt{2}$)

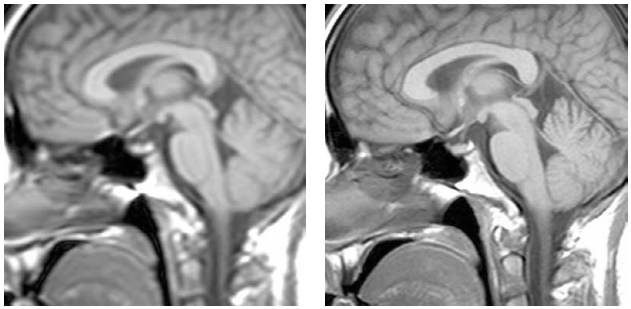
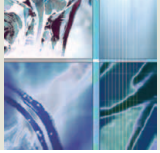


Image comparison: Matrix 128 (left) low resolution and better SNR, Matrix 256 (right) higher resolution and decreased SNR.



The image field

The size of the measurement matrix determines spatial resolution. The in-plane resolution determines the actual, two-dimensional resolution in the image. The pixel size is instrumental for in-plane resolution. The pixel size results from the measurement matrix selected and the image field.

What is the image field?

The IMAGE FIELD OR FIELD OF VIEW, abbreviated as FOV, is the basic size of the slice section to be measured (in mm). In short, the FOV determines what you can see in the MR image.

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

Let's first take a look at a square FOV.

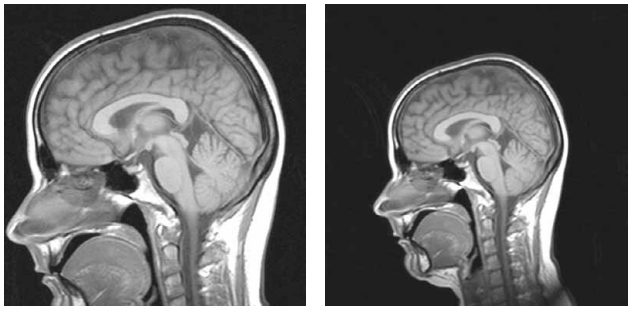


Image comparison: FOV = 230 mm (left), FOV = 330 mm (right) is unnecessarily large.

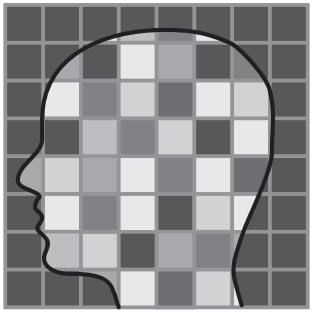
Pixel size and in-plane resolution

The smaller the field of view at a fixed matrix size, the higher the in-plane resolution. The number of pixels per in-plane unit increases while the pixels as such decrease in size. Vice versa, at a given matrix size and at a larger FOV, pixels are downright "blown up". Resolution is reduced accordingly.

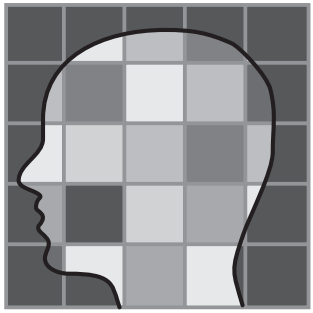
$$\text{Pixel size} = \frac{\text{FOV}}{\text{Matrix size}}$$

Smaller pixels mean an improved in-plane resolution

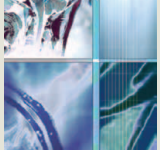
FOV (mm)	Matrix size	Pixel (mm)
256	256×256	1.0×1.0
256	128×128	2.0×2.0
128	128×128	1.0×1.0



FOV 256 mm
Matrix 256×256



FOV 350 mm
Matrix 256×256



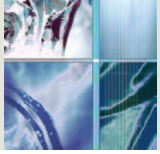
Reduced to the essentials

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

Changing these measurement parameters has a number of effects. For this reason, the most optimal solution is a compromise—primarily between image quality and measurement time.

Summary of the effects of the parameters:

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

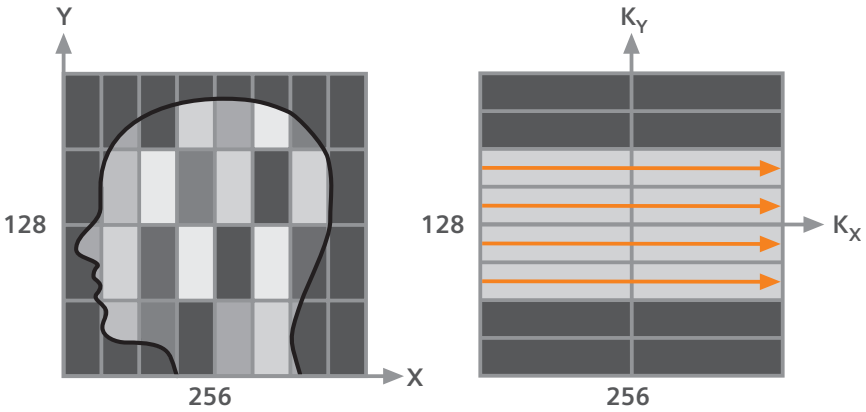


Measuring faster

Faster measurements are advantageous to both you and the patient. We can contribute to it via the measurement matrix and field of view. What are the effects on SNR and resolution?

Rectangular matrix and rectangular pixels

So far we have shown you how MR images are computed from square measurement matrices. To accelerate the measurement, you are able to select a reduced measurement matrix with a lower phase resolution, e.g., in place of 256×256 you'll select 128×256. The pixels are now rectangular.



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

A phase resolution cut in half (e.g., 128) corresponds to half the number of phase-encoding steps. The measurement time is cut in half.

Filling the k-space

The measurement matrix selected by us is rectangular, although the k-space has to be a square, always. Signals are acquired only for the central raw data lines. The missing outer rows in the k-space are filled with zeroes.

Why is this working? Fine structures are shown in the outer regions of the k-space. The central rows provide the important contrast. When reconstructing an image from k-space, the image pixels are interpolated in the phase-encoding direction.

How is the image quality? Image resolution is reduced with the phase resolution of the measurement matrix (e.g., phase resolution is reduced by 50%, image resolution is reduced by 50% in this direction). Since the voxels are larger, SNR is improved.

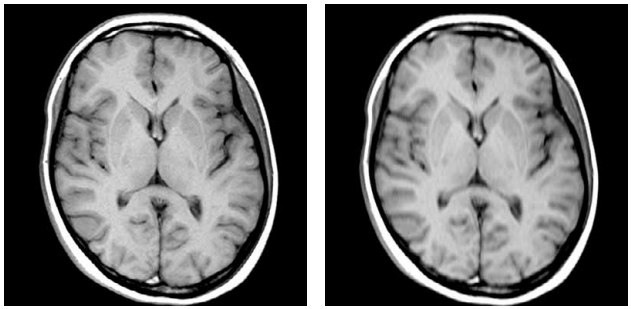


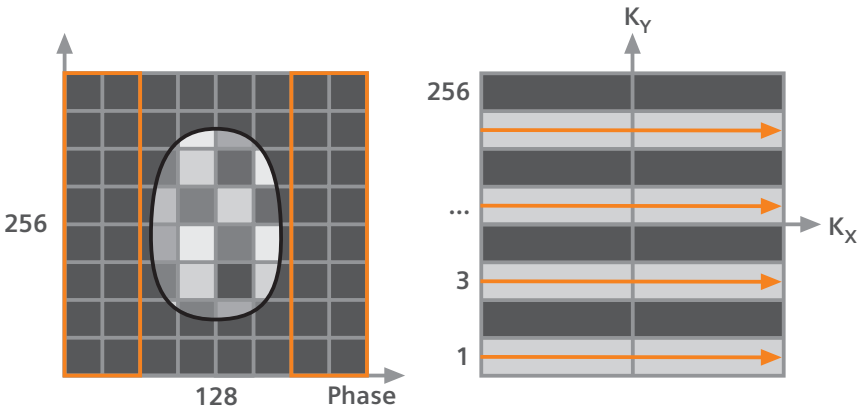
Image comparison: Phase resolution 100% (left) and 50% (right).

Measuring faster with a rectangular field of view

We can do more than just reduce the measurement matrix. The field of view can be rectangular as well—either in addition to or as an alternative. The image gets smaller in the phase-encoding direction as well. As we have seen: This is how we reduce the measurement time.

Rectangular FOV and square pixels

When the object to be measured does not fill a square image, we can select a rectangular field of view (FOV). If we cut the FOV in half in the phase-encoding direction, we only need half as many phase-encoding steps. As a result, the scan distance in the k-space is reduced: only every other row is filled with raw data, the others contain zeroes only.



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

Measuring faster at the same resolution

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

SNR decreases.

Usually, the decreasing SNR barely affects the quality of MR images. For this reason, a rectangular FOV is a good choice for accelerating image acquisition.

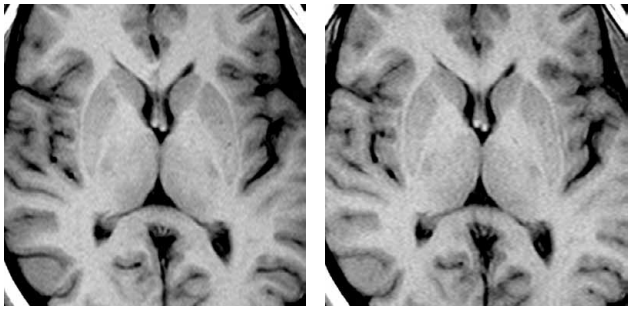
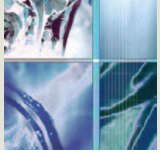


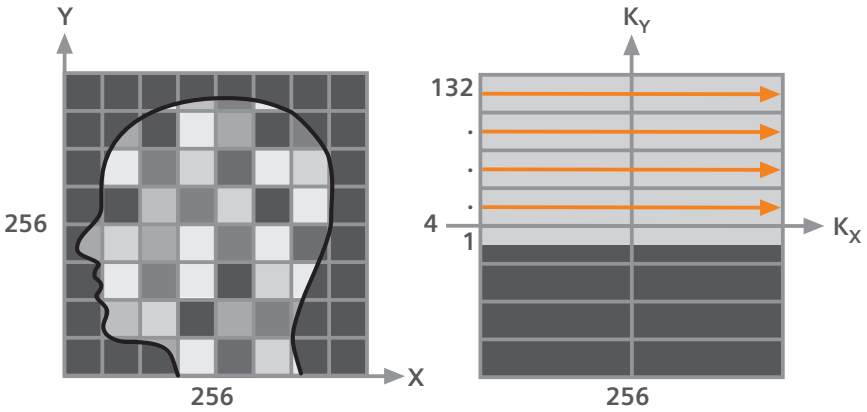
Image comparison: FOV Phase 100 % (left) and 50 % (right).



Measuring faster due to the symmetry of the k-space

The k-space is symmetric. With respect to image information, it is therefore sufficient to fill it in part only. Missing information is reconstructed symmetrically.

Half the k-space and square pixels
Using Half Fourier technique, only half the k-space is filled with data in the phase-encoding direction.



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

As a result, the measurement is almost twice as fast.

More than half the k-space

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

How is the image quality? Because of the same voxel size, resolution is of the same level of quality. SNR decreases. In most cases, there is a barely discernable difference between images with and without Half-Fourier.

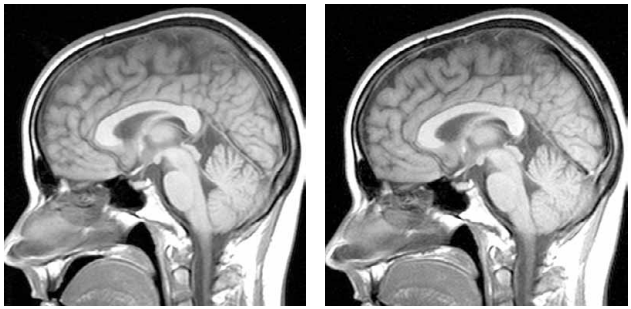
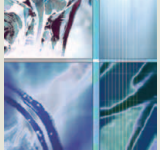


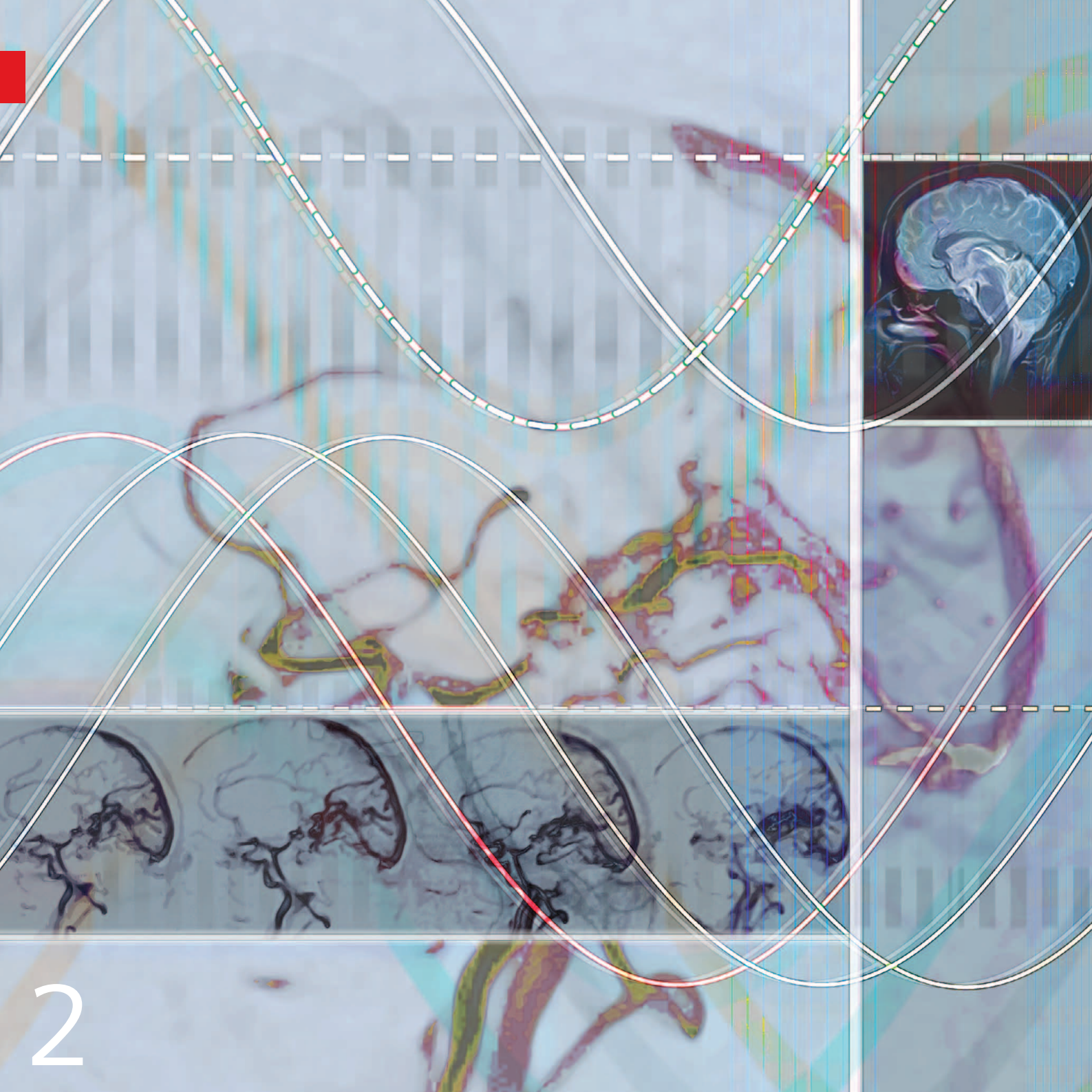
Image comparison: Normal (left) and Half-Fourier (right).



Summary

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

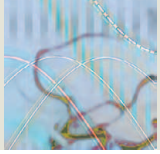
	SNR	Resolution
Field of view 100% Phase resolution 50%	Better	Less
Field of View 50% Phase resolution 50%	Less	Unchanged
Half the k-space	Less	Unchanged



Spins in flux and motion

In the first volume of *Magnets, Spins, and Resonances* we limited ourselves to spatially fixed spins. In reality, however, many of the spins in the body are in motion (blood flow, CSF). These either are going to be displayed or suppressed as interfering flow effects.

Angiography and cardio-vascular imaging are examples for MR-applications that take advantage of the effects of flowing spins. Time-of-Flight and phase contrast are two techniques that may be applied for this purpose.



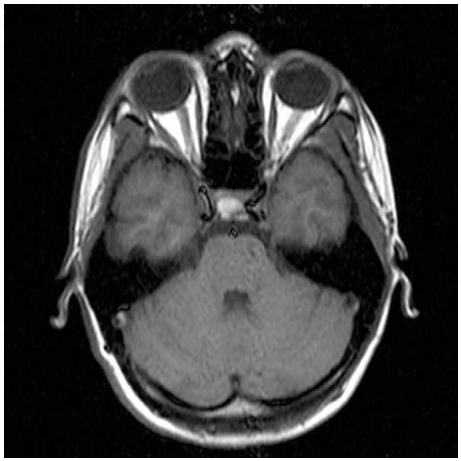
Time-of-Flight (ToF) — The flow through the excitation slice

Using special techniques, MR images are sensitized to flowing spins. In this way, blood may be deliniated as high in signal or low in signal with respect to the surrounding stationary tissue. Both displays are based on the ToF-effect, the short dwell time of flowing spins in the slice.

Flow in T_1 contrast

Flow means blood flow, liquor, etc. Since the flow effects are the same independent of the respective body fluid, we are going to limit our example to blood flow.

Blood has a relatively long T_1 relaxation. A standard T_1-weighted image shows blood vessels as dark structures. The surrounding tissue with shorter T_1 relaxation is shown as bright.

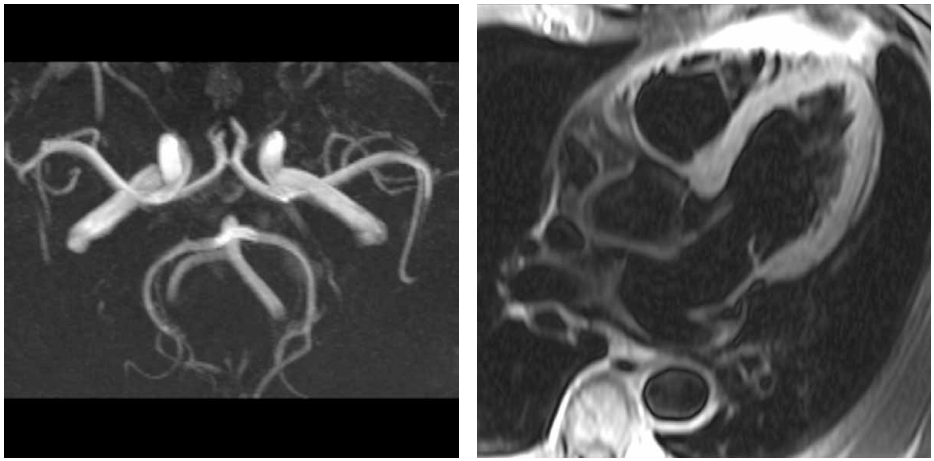


Standard T_1-weighted image.

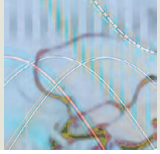
The contrast is sensitized to flow

We would like to take advantage of the flow effects. Flow-sensitive pulse sequences allow us to display blood as very bright or as almost black.

In Angiography vessels are displayed by using Maximum Intensity Projection, also known as MIP. Maximum intensity projections are computed from 3-D or multiple-slice measurements and combined into MIP series.



Blood shown with high signal (to the left, head image, MIP display) and suppressed signal (to the right, heart image, single slice).

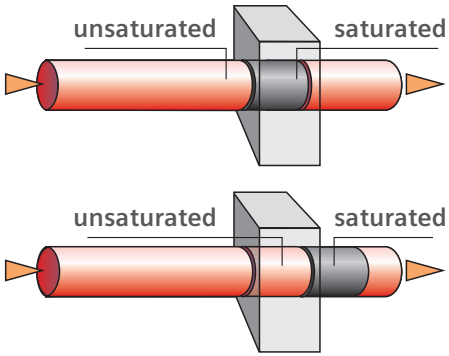


Bright blood through inflow effect

The inflow effect is the basis for the ToF technique : As compared to stationary tissue spins, blood spins flowing into the slice are only briefly affected by the pulse sequence. The velocity of the blood flow determines how quickly they are replaced by subsequent spins flowing in. This also affects the brightness of the blood display.

Saturating stationary spins ...

The spins are saturated by fast excitation pulses in the slice stimulated. Outside the slice, the spins are *not* stimulated. As soon as they flow into the slice, the excitation pulse affects very high magnetization. The subsequent data acquisition displays the unsaturated blood as bright, and the surrounding tissue as low in signal.



The excited slice is saturated via a short repetition time TR. It is considerably shorter than the repetition theoretically *required* for signal recovery. This prevents recovery of the longitudinal magnetization in the slice.

At first, the inflowing blood spins generate a strong signal after the excitation pulse. Their saturation increases, if they are exposed several times to the excitation pulse.

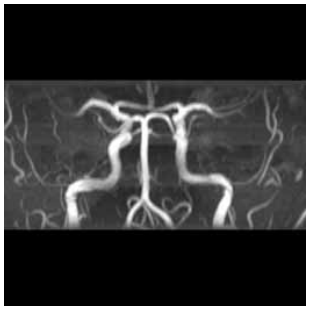
... and blood is displayed bright

In case of optimal pulse timing, the vessel spins of the saturated slice are replaced by fresh, unsaturated spins prior to a new excitation pulse followed by data acquisition. Through the amplification of inflow, blood in the vessel is shown at maximum brightness: BRIGHT BLOOD.

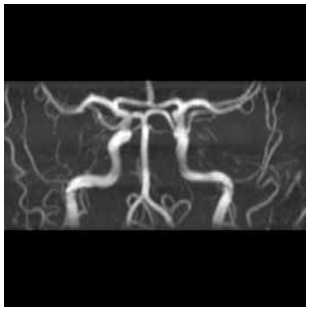
When does the blood signal reach this maximum level? We can compute the associated repetition time TR for a given slice thickness and blood velocity:

$$TR = \frac{\text{Slice thickness}}{\text{Blood velocity}}$$

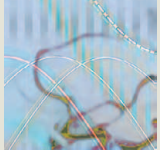
Example: At a slice thickness of 5 mm and a flow velocity of 12.5 cm/s, we obtain an optimally bright blood signal at a TR of 40 ms.



TR is too large: Low inflow amplification.



Optimal TR: Maximum inflow amplification. The display of distal vessels is enhanced.



Slice orientation and presaturation

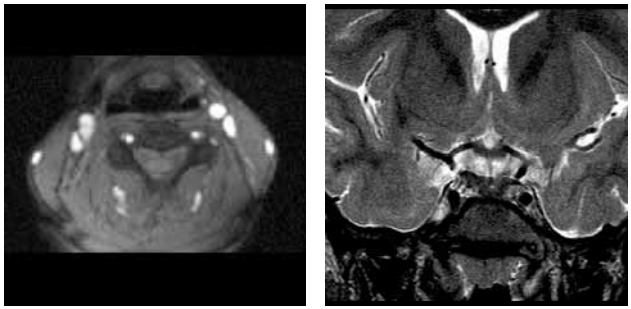
By selecting TR correctly, we obtain optimal contrast between blood and tissue. The alignment of the excitation slice with respect to the blood vessel can be optimized as well. To prevent display of a certain blood vessel, the signal of the vessel spins may be suppressed as well via presaturation.

Optimal vessel course

Let's look at the course of a blood vessel through the excitation slice (through-plane). If the slice is located orthogonally to the vessel, we have just a short vascular section within the excitation slice. The spins have only a short dwell time in the slice. They are continuously replaced by new inflowing, unsaturated spins.

When the blood vessel lies longitudinally to the excitation slice (in-plane), the spins remain much longer in the slice. They are saturated more and more by the repeated excitation pulses. Their signal decreases.

> The blood vessels are optimally displayed when the excitation slice and the blood vessel are located orthogonally to one another.

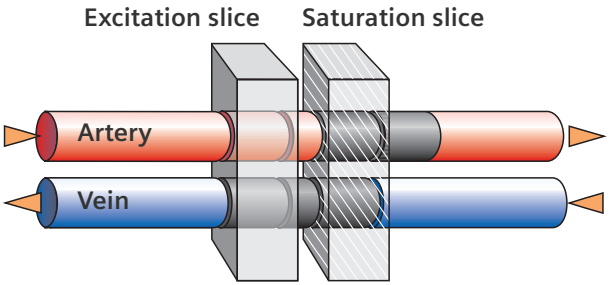


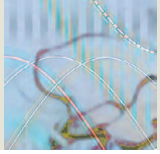
Excitation slice, orthogonal (left), used primarily for displaying carotids, and excitation slice longitudinally to blood flow (right).

Artery or vein?

In most body regions arterial and venous blood flow in opposite direction. As one of the advantages, just one of the two blood flows is displayed in bright.

Let's assume we would like to show the arterial flow and suppress all venous vessels. For this reason, the vein in the excitation slice may contain saturated spins only. On the side of *venous* inflow, we position a parallel saturation slice ➔ page 58 in front of the excitation slice. Venous spins flowing through it do *not* contribute to the signal during subsequent inflow into the excitation slice. Only the spins in the artery are displayed as bright in the image.





Blood is displayed as dark

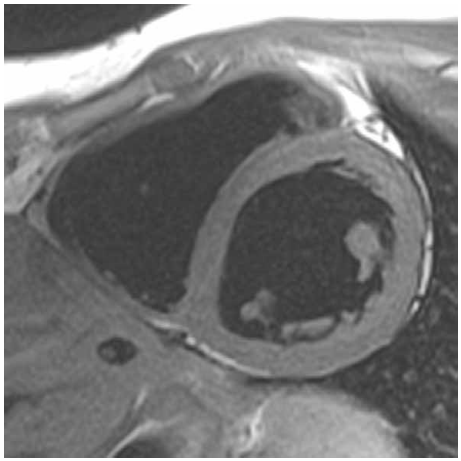
In certain cases, complete nulling of the blood signal is useful.
The dark blood effect used for this purpose is frequently used in cardio-vascular imaging for morphological displays of the heart. How does this work?

The dark blood method

The first 180° pulse *inverts* the blood and tissue signal within *and* outside the excitation slice.
In cardiac imaging this means an inverted signal across the entire thorax.

A subsequent 180° pulse affects the excitation slice only (e.g. the slice through the heart). The signal is now reinverted.

The blood subsequently flowing into the slice was inverted by the first 180° pulse. In case it flows through the slice during the zero-crossing of its magnetization and data are acquired during this time, only the adjacent tissue will generate a signal. The blood itself is shown in black.

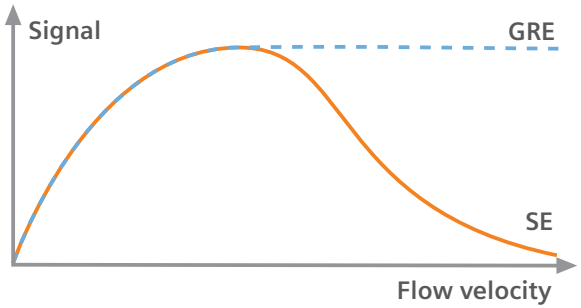


Dark blood imaging of the heart.

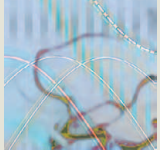
DISCUSSION

Spin echo and wash-out effect

Until now we have looked at the spins independent of the pulse sequence. With spin echo sequences and high blood velocities, the signal is attenuated without a preceding inversion. As long as the flowing spins are *not* exposed by the two pulses of the spin echo sequence, they do not contribute to the signal. The faster the blow flow, the fewer spins are exposed to the 180° pulse in the slice. The signal grows weaker. This is known as the wash-out effect. We have a complete signal void if all of the spins excited by the 90° pulse have flown out as soon as the 180° pulse was applied. This is called dark blood.



Gradient echo sequences do not have a wash-out effect (excitation pulses 180° and 90° are omitted). This is why they find preferred use in MR-Angiography. Additional advantages of gradient echo sequences are their short repetition times. This allows for better suppression of the signal from stationary tissue as well as faster measurements.

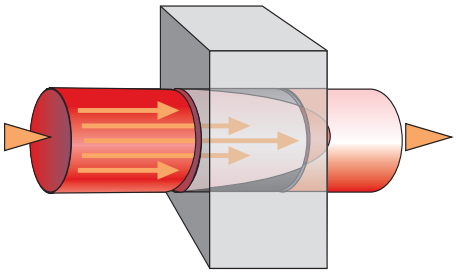


Laminar flow and turbulences

Up to now we have only described the most optimal conditions for blood flowing through the measurement slice. In practical application, however, different effects influence the signal of blood flow. In part these effects are caused by the phase shift of the flowing spins.

Signal attenuation through laminar flow

Let us look at the flow in blood vessels. Most of the time we can observe a decrease in velocity from the center of the vessel to the vessel wall. This is known as the LAMINAR FLOW.



The dwell time of the spins in the slice decreases from the center to the edge of the vessel. As a result the number of fresh, unsatured spins in the flow direction is decreasing in the slice.

With thick slices and/or slice stacks, this may lead to signal attenuation along the course of the slice(s).

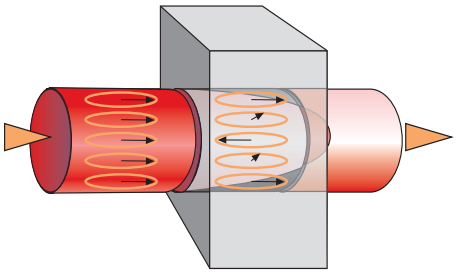
Laminar flow and spin dephasing

We are now comparing the signal of a laminar and non-laminar flow in a gradient echo sequence. During a laminar flow, the signal is weaker under the same measurement conditions.

In this case, the phase of the spins comes into play. Spins moving along a gradient are subject to phase shift ➔ page 44.

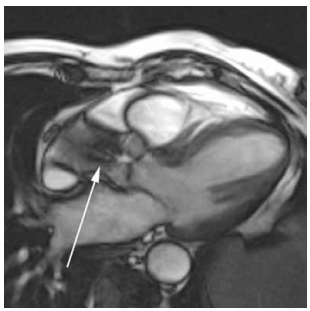
> The faster a spin is moving along a gradient, the stronger the phase shift.

The fast flowing spins in the center of the blood vessel experience a stronger phase shift than the slower flowing spins at the edge of the vessel. The phase coherence between the spins in the blood is lost, the spins are being dephased. The signal is reduced.

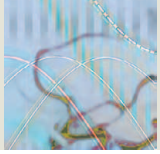


Turbulences and jet effect

A turbulent flow behind a vessel stenosis also leads to signal reduction or even signal void. This is known as the JET EFFECT. In this case the phase coherence of the spins is lost as well, they are being dephased.



Jet effect caused by turbulent flow.



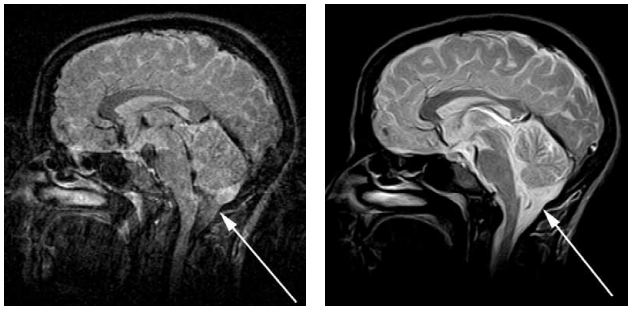
Signal loss and flow compensation

How can we minimize or even eliminate interferences caused by dephased spins? This isn't just of interest for displaying flow. Images of body regions with a high volume of flowing spins benefit from this as well.

Signal loss through flow

Flowing spins that are being dephased lead to signal loss and incorrect encoding. This is interruptive when imaging vessels. Even the images taken of entire body regions may be adversely affected.

In the area of the thoracic spine, cervical spine and head, the influence of moving spins in blood or liquor on the MR image is especially noticeable. You have to compensate for signal losses in order to obtain optimal image results.

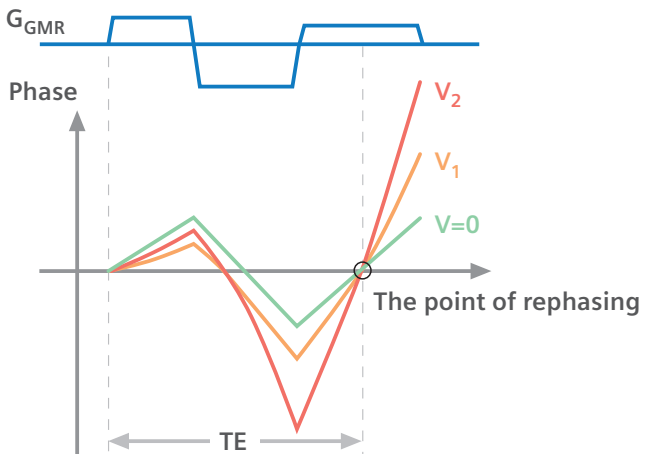


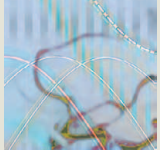
Images without (left) and with flow compensation (right). In the left image, flow leads to a region of artifacts and signal reduction in liquor (arrow) and blood vessels. In the right image, flow compensation reduces the artifacts and enables a high signal of liquor and blood vessels.

Flow compensation via GMR

To cancel signal loss and incorrect encoding through spin motion, moving and unmoving spins need to be rephased. This is possible by using Gradient Motion Rephasing (GMR).

Additional gradient pulses are switched in suitable size and time duration. They compensate for the phase shift of stationary spins (v = 0) and spins flowing at different levels of velocity (v_1, v_2) and restore phase coherence. The strength of the signal at the point of rephasing is the same as prior to dephasing.





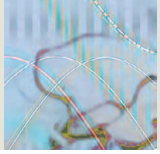
Reduced to the essentials

The Time-of-Flight technique clearly delineates flowing spins from stationary spins. For this purpose, the stationary spins in the slice are saturated. Inflowing unsaturated spins provide for a stronger signal. This allows us to display the course of vessels (angiography).

Flowing spins may also generate interfering flow artifacts. The rephasing gradient (GMR) is switched for flow compensation.

With laminar flow, the signal is attenuated through spin dephasing along the slices. Turbulent flow may lead to full signal nulling (jet effect).

The phase-contrast technique

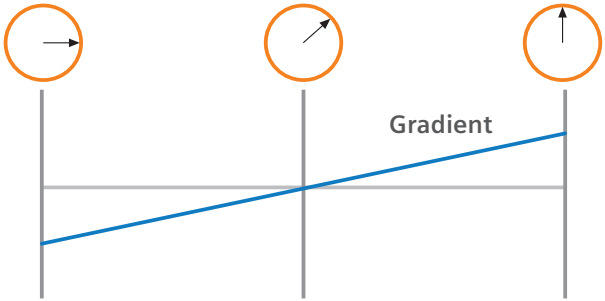


The phase-contrast technique

To display flow, the time-of-flight technique takes advantage of the dwell time of flowing spins in the excitation slice. Let's take a look at the technique that works exclusively with the phases of spins and their phase shift. We call this the phase-contrast technique.

Moving spins and gradients

We have determined that the phase of a spin shifts when the spin moves along a gradient. When using Time-of-Flight, this leads to interferences that have to be compensated for by additional gradients (GMR technique).



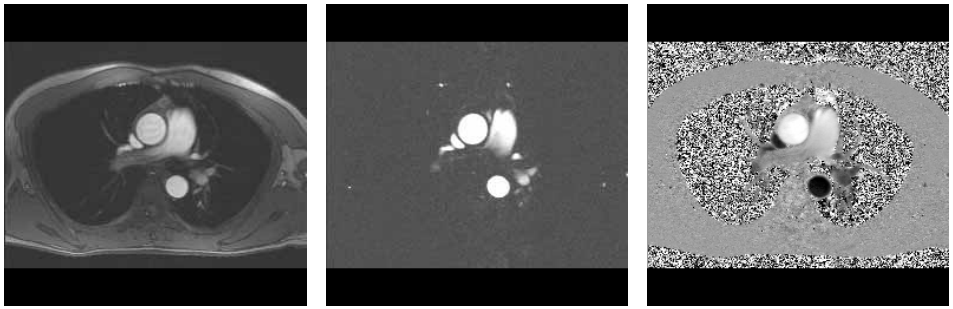
We take advantage of this effect when using the phase-contrast technique. On the basis of the phase shift, we would like to determine whether and how the spin is moving. What we considered an interference before, we are now turning into a measurement principle. And we are making the best of a simple rule:

> The phase shift of flowing spin increases linearly with a constant gradient field and increasing flow velocity.

This is how phase difference is being determined

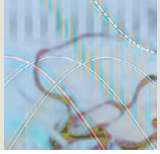
Phase-contrast techniques perform reference measurements where all spin phases are in the same position. We obtain this by acquiring a *flow-compensated* image. Flowing spins are shown the same way as stationary spins.

Subsequently, a gradient pulse is switched in one direction. The phase of the flowing spin changes. The *flow-encoded* data are generated in a subsequent measurement that is not flow-compensated.



Flow-compensated image (left), flow-encoded image (center), and phase-contrast image of a transverse thoracic image (right).

The phase-contrast image is generated from the difference between the two phase informations (T_2-weighted). There is no display of anatomical information. Instead the pixels show the phase difference of the spins. Bright pixels represent a high flow velocity in the positive direction, dark pixels represent a high flow velocity in the opposite (negative) direction. The mean grey value represents a flow velocity of zero, that is, stationary tissue.



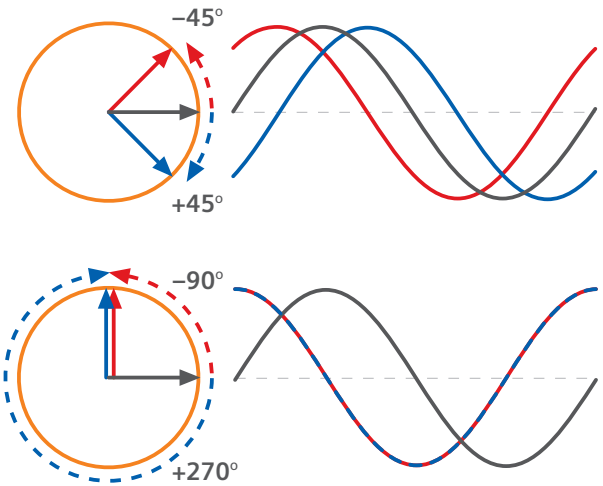
Phase shifts, phase inversion and flow sensitivity

Until now we have talked about phase shift as if it were a magnitude without an algebraic sign. However, the algebraic sign in front of the phase shift is of importance.

Phase shift and phase inversion

Phase shifts up to ±180° have an unambiguous algebraic sign and can be correctly displayed in phase-contrast images. The algebraic sign of a phase difference becomes problematic, when the phase shift exceeds ±180°.

Let's use a phase shift of +270° as our example. In the sinus display this corresponds exactly to a phase shift of –90°. And it is also recorded as this smaller value. A PHASE INVERSION occurs: The +270°-phase shift is shown as a –90°-phase shift.

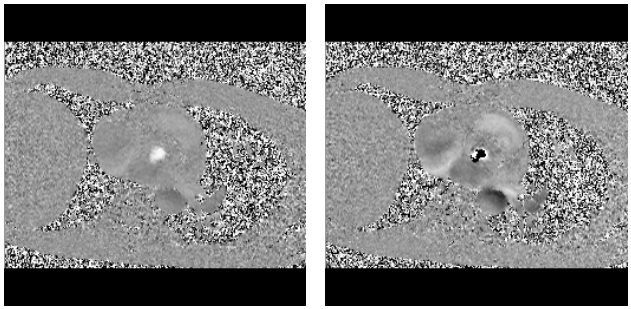


Regulating flow sensitivity

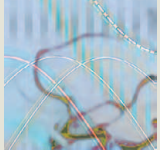
How can we prevent phase inversions? The magnitude of the occurring phase differences can be controlled via the gradient where the spins are flowing. Via its parameters, for example, strength and duration, the phase difference can be increased or reduced at the same flow velocity. The gradient has a parameter-dependent FLOW SENSITIVITY (velocity encoding, venc).

As long as blood velocity remains within the range of flow sensitivity, the phase differences are not going to exceed the limit value of ±180°. In case the phase differences are too small because of high flow sensitivity, they will disappear in the signal-to-noise ratio.

Phase inversions occur if the spin velocity exceeds the flow sensitivity of the pulse sequence. Incorrect grey values are displayed in the phase-contrast image.



Phase-contrast image without (left) and with phase inversions (right).



Magnitude images, phase magnitude of flow

The direction of flow is frequently known or not important for diagnostic purposes. In this case, you can select a different display for image information: the phase magnitude.

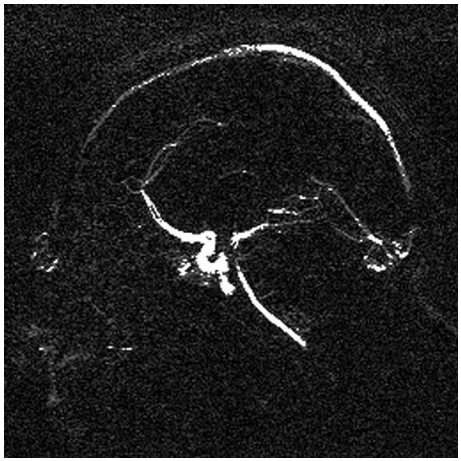
Displaying the magnitude of flow sensitivity

If information regarding the direction of flow is not important, the algebraic sign for the phase shift is equally unimportant. It will suffice to know the magnitude of the phase shift or phase difference.

This magnitude is visualized in the MAGNITUDE IMAGE. Stationary tissue is shown in black.

> The higher the flow velocity, the brighter the grey value of the pixel in the magnitude image.

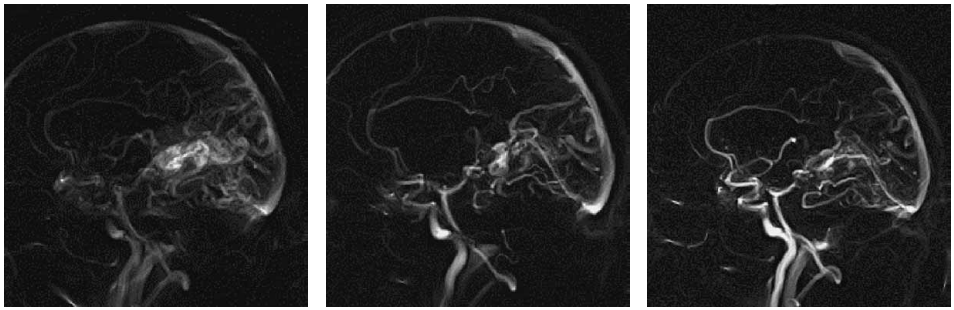
The same ±180° limitation applies for displaying the phase difference in magnitude images.



Magnitude image.

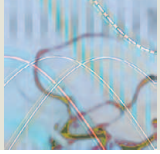
Displaying spatial flow sensitivity

If gradients are switched into the three orthogonal directions, three magnitude images of the same slice are generated, however, with different contents. Spin movement along the gradient is shown as bright pixels, spin movement orthogonal to the gradient is *not* shown (black pixels).



Magnitude sum image with venc = 10 cm/s (left), 30 cm/s (center) and 60 cm/s (right).

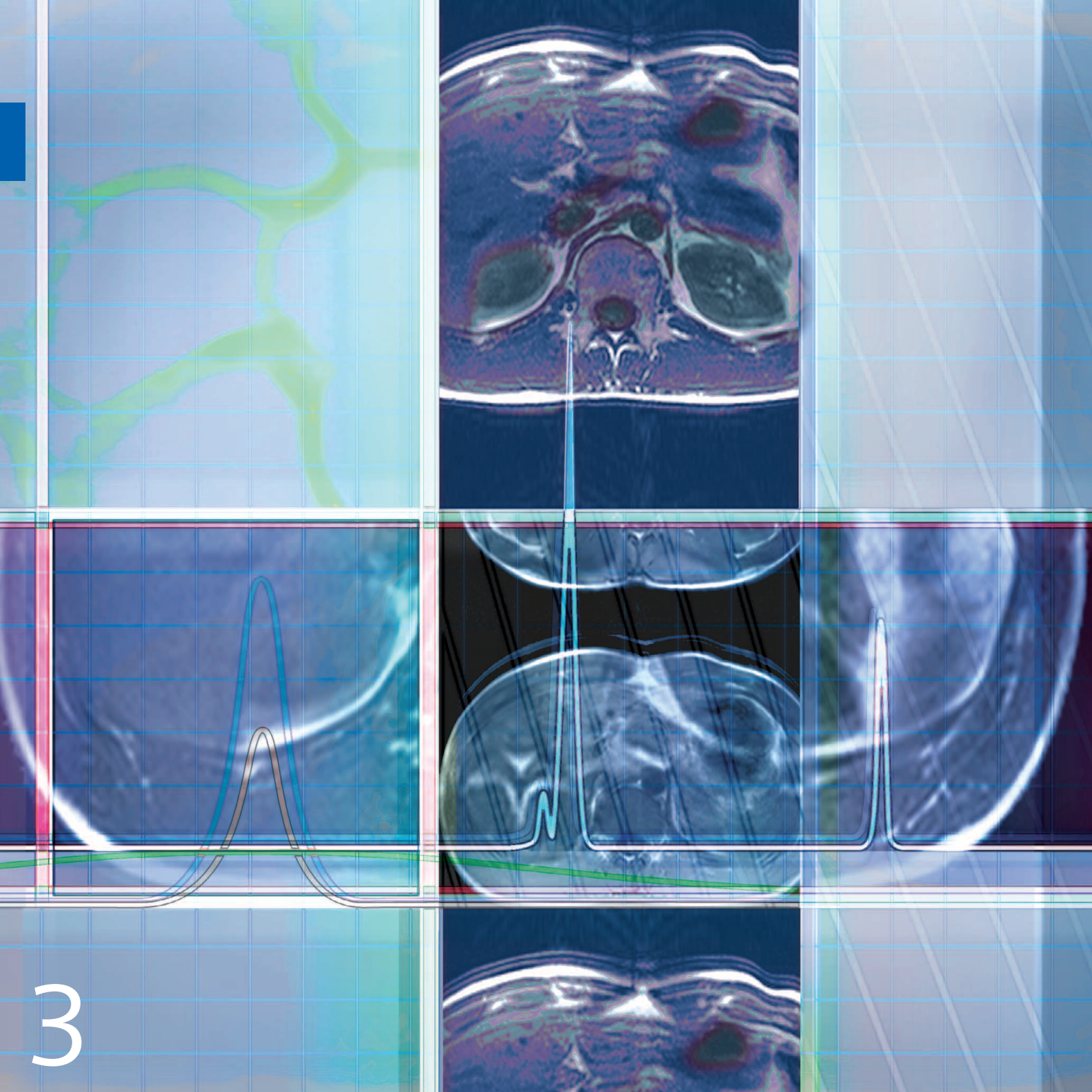
As compared to phase-contrast images, we can add up the three magnitude images of the orthogonal directions. We obtain an image that shows flow in all directions, the MAGNITUDE SUM.



Summary

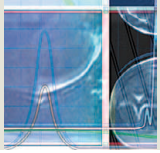
Flowing spins influence the MR signal through inflow into a slice. Spins within a slice that move along a gradient change their signal as well.

Time-of-Flight effect and phase shifts provide the possibility of visualizing flowing spins either via T_1-contrast (ToF) or phase contrast.



Saturation and chemical shift

When we talk about good contrast in MR imaging, we mean the contrast obtained in the image regions or tissue of interest. Frequently these areas are irradiated by high signal portions from other image regions or from tissue that is not of interest. Saturation is helpful in dispaying the desired contrast. In this chapter we are going to describe the various saturation techniques, their common features, their differences as well as their various applications based on individual examples.



Spatial saturation

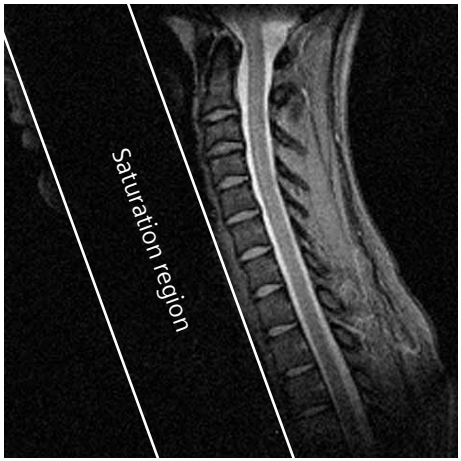
Saturation does not equal saturation. Do we want to saturate the tissue of an entire region, or just the spins that flow into the region? Then we are going to select the most suitable technique from the list of spatial saturation techniques. Or does the signal of an individual tissue interfere that occurs throughout the region? In this case, selective saturation is of practical use. However, we think it would be best to start with spatial saturation.

Spatial saturation techniques

Spatial saturation includes three different kinds of application:

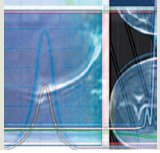
- Presaturation
- Parallel saturation
- Traveling saturation

For presaturation and parallel saturation, three defined regions are selected. Traveling saturation, on the other hand, travels with the image slice of a slice stack.



Example of spatial saturation.

Tissue-selective saturation



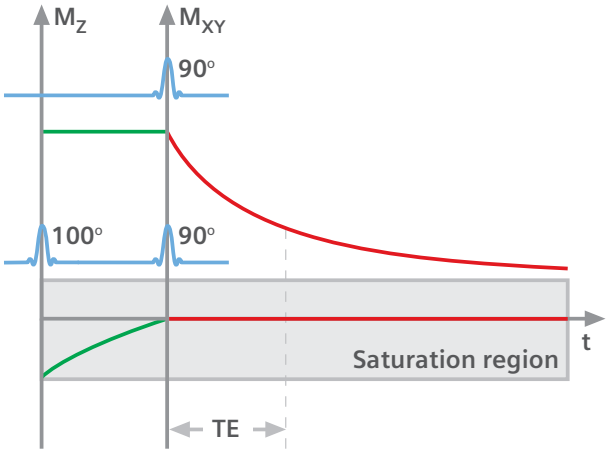
Spatial presaturation

Using one or several saturation pulses, the signal of undesired tissue may be reduced. Since the pulses are switched in front of the actual pulse sequence, we are talking about *pre*saturation.

Presaturation pulses before the sequence ...

A 100°-saturation pulse flips the entire magnetization M_z within the saturation region under the transverse plane M_{xy}. As soon as the magnetization of the saturation region reaches zero, measurement with the excitation pulse begins.

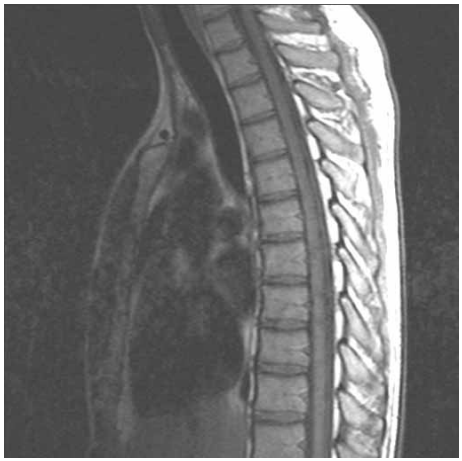
Since the time interval between saturation and excitation pulse is much shorter than T_1 of the tissue, the longitudinal magnetization in the saturation region can relax very little. Magnetization is low. This area generates a very low signal and appears dark in the MR image.



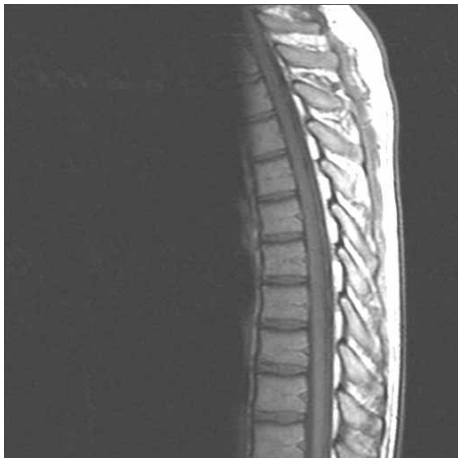
... eliminates artifacts caused by motion or flow

The application range for presaturation is demonstrated on the basis of the following example: Artifacts caused by cardiac motion or blood flow may be included in the sagittal image of the thoracic spine.

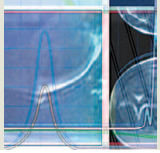
Interferences of this kind occur only because the area of the artifact actually contributes to the overall signal. If magnetization of this region is weak, the small signal component cannot lead to interferences in other regions.



Without presaturation, artifacts appear in the image—especially in the vertebral bodies.



With presaturation artifacts are reduced.



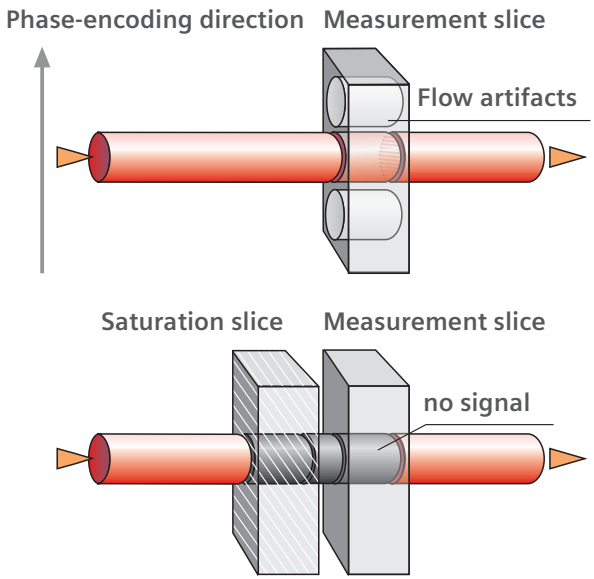
Parallel saturation

Parallel saturation is basically a variation of presaturation. Again, the saturation pulse is switched in front of the actual pulse sequence.

Saturation regions parallel to the slice ...

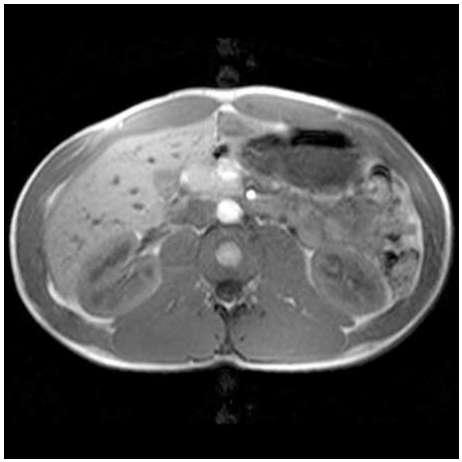
When imaging blood vessels, flow artifacts in the form of ghosting may occur in the phase-encoding direction. Parallel saturation eliminates the source for ghosting.

The saturation region is not in but rather outside the imaging slice. Parallel alignment to the imaging slice affects the signal void of flowing spins. Blood flowing from the saturation region into the imaging slice does not generate a signal. This technique is used for Time-of-Flight ➔ page 35 as well. Arteries and veins can be imaged separately.

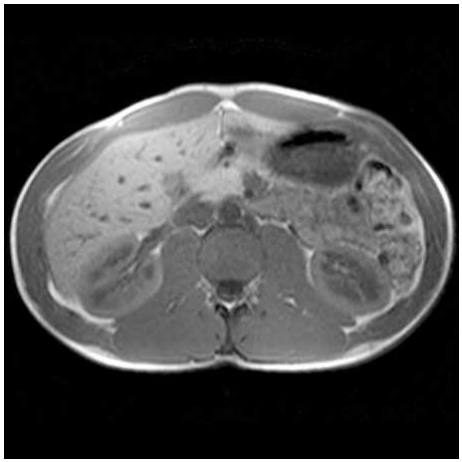


... eliminate flow artifacts

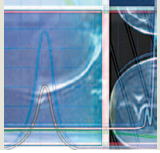
To avoid flow artifacts, a parallel saturation slice is positioned in front and in back of the slice to be imaged. In this way, both arterial and venous blood are saturated. Flow artifacts are suppressed.



Without parallel saturation, flow artifacts occur in the phase-encoding direction. The artifacts are caused by arterial pulsation of the aorta.



Flow artifacts are suppressed with parallel saturation



Traveling saturation

When acquiring an image stack, the distance to the stationary parallel saturation slice increases with each slice. The flowing spins in the blood vessels can relax between saturation and image slice. Their signal strength is increasing again.

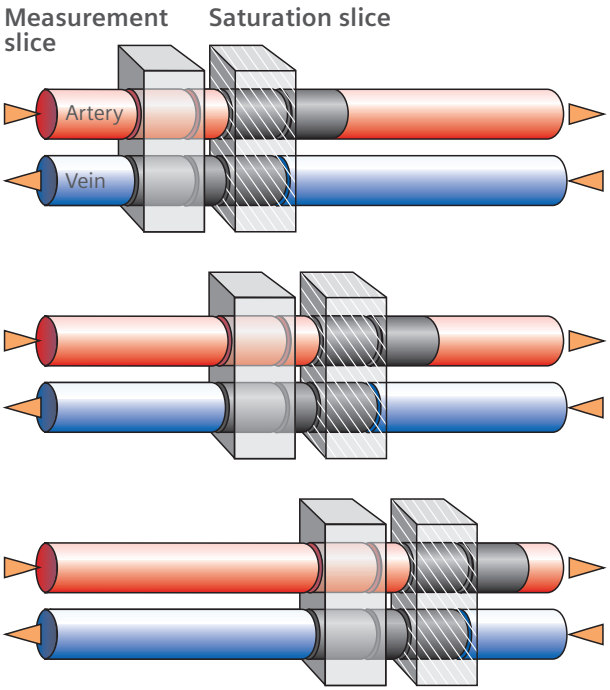
The saturation travels with the image slice

The increasing signal strength of the flowing spins may lead to ghosting again in the last slices of the stack. This is prevented by the traveling sat.

The parallel saturation slice is no longer stationary. Instead it is shifted together with the image slice.

As compared to stationary parallel saturation, traveling saturation slices are positioned at one side of the image slice only. Otherwise, the entire subsequent image slice could be already presaturated.

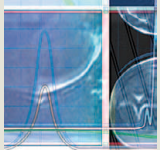
Travel sats are only possible with sequential slice sequences. They may not be used for interleaved multi-slice measurements. The meassurement software automatically eliminates this choice.



Reduced to the essentials

Spatial saturation methods are well suited for excluding motion and flow artifacts.

The saturation pulse always precedes the excitation pulse. These techniques differ in location (within and outside the slice) and in a possible change in position (stationary or traveling)



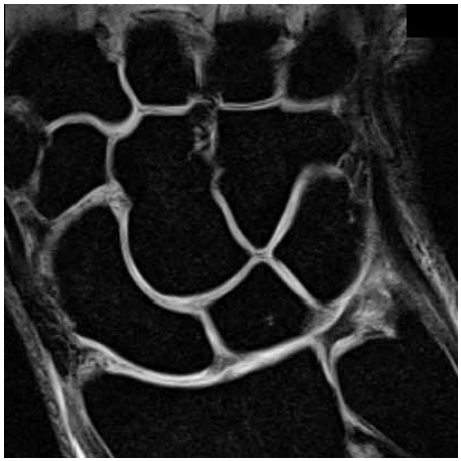
Tissue-selective saturation

Since we have covered spatial saturation, we would now like to address selective saturation. In connection with saturation, selective means signal suppression of a specific tissue or fluid.

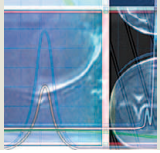
Techniques of selective saturation

We are describing three areas of tissue-selective saturation:

- Dark Fluid and STIR, as a function of relaxation time
- Fat/water saturation, frequency-selective
- Magnetization transfer (MTC)



Example of selective saturation: wrist with frequency-selective fat saturation.



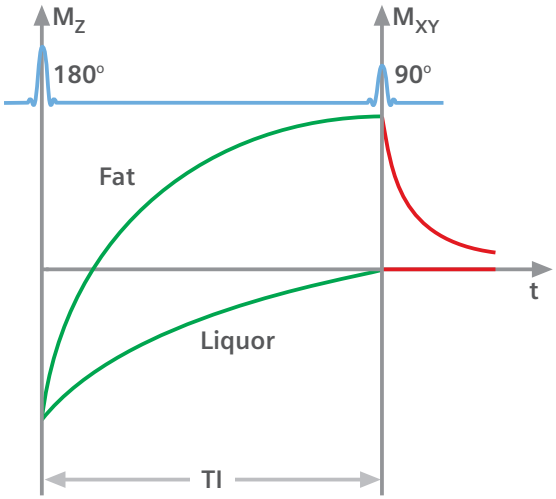
Dark Fluid Technique

The Dark Fluid Technique saturates liquor. Since this technique works with an inversion pulse, it is known as FLAIR (Fluid Attenuated Inversion Recovery).

Saturation with an IR pulse ...

We flip the longitudinal magnetization by 180° via an IR pulse and wait until the liquor signal passes through the zero crossing of the magnetization.

At this point, we switch the stimulation pulse in the slice. Since the liquor signal is at zero, it is not stimulated at this time. It does not generate a signal. Liquor is displayed in black.

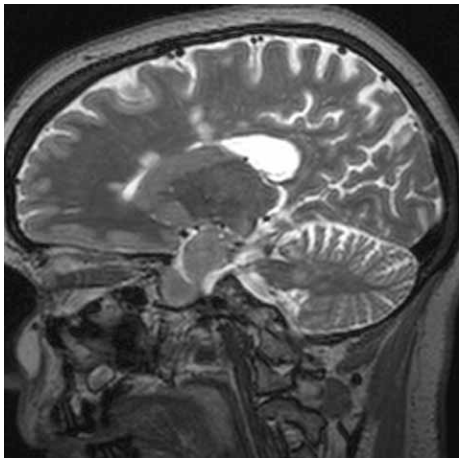


... eliminates the liquor signal

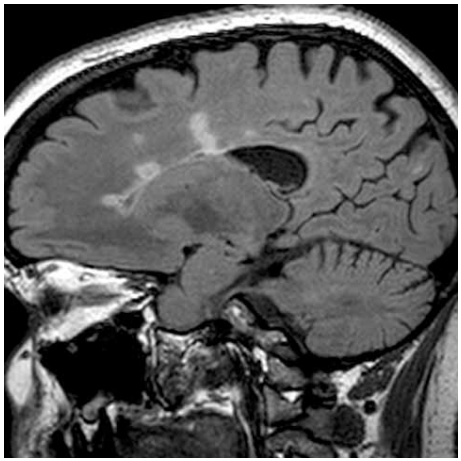
The liquor signal is very strong during non-saturated imaging and may superimpose lesions. Through Dark Fluid technique, the signal portion of the lesion is more visible.

Using the same technique for STIR (Short TI Inversion Recovery), the fat signal is suppressed. After a short TI, the excitation pulse arrives at the zero-crossing of the fat signal.

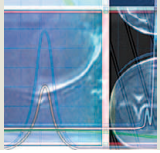
A detailed description of STIR is included in the first volume *Magnets, Spins and Resonances*.



Turbo spin echo image.



Dark Fluid image.



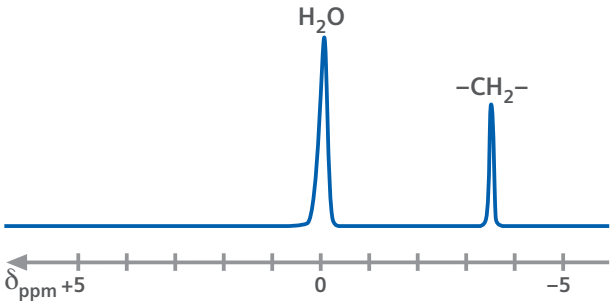
Fat/water saturation

The MR signal is the sum of signals from water and fat protons. When we saturate the signal portion of one these two proton types, we obtain a clearer picture of the other proton group.

Chemical shift ...

In water and fat molecules, the hydrogen atoms are bound to different positions. This affects the strength of the magnetic field experienced by a fat or water proton. A weaker magnetic field affects fat protons than water protons. For this reason, the resonance frequency of the fat proton lies somewhat below that of the water proton. Hydrogen nuclei within a molecule provide *different* resonance lines for fat and water. This shift in resonance frequencies is known as CHEMICAL SHIFT. It is shown by the shift in associated resonance lines in the spectrum measured.

When we transmit a frequency-selective saturation pulse, only the protons with the respective resonance frequency are saturated. Their signal portion is suppressed.



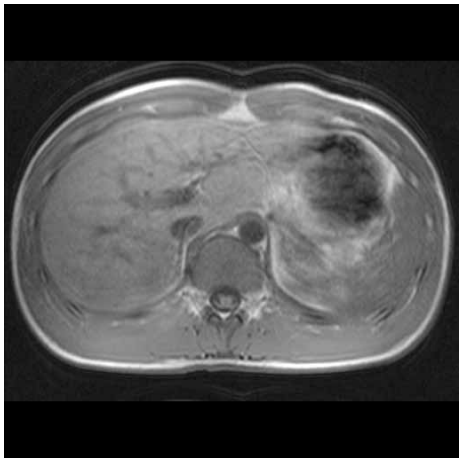
Chemical shift of 3.4 ppm for water and methyl group ($-CH_2-$), main component of fat. The chemical shift is expressed in δ_{ppm} (ppm = parts per million). $\delta_{ppm} = -3.4$ means that the frequency of the methyl group is reduced by 3.4 of a millionth as compared to water. In spectral displays common to MR, the frequency axis is oriented from right to left.

... and elimination of the fat signal

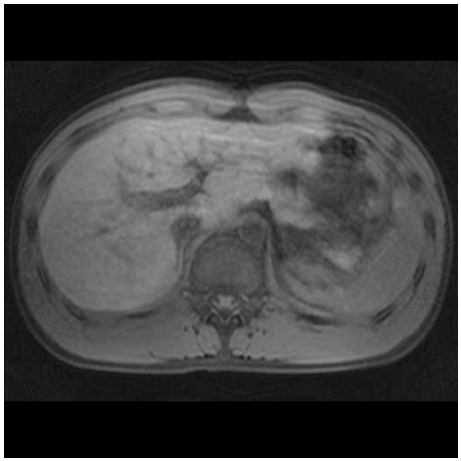
Fat is represented by a high signal in images with almost all contrasts. This leads to a loss in contrast between the tissues of interest. Also, an increase in motion artifacts may be present.

These effects are eliminated through the frequency-selective saturation of fat protons.

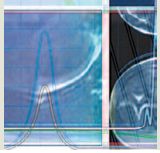
Accordingly, the water signal can be suppressed with water saturation. This is utilized especially in the area of MR spectrocospy.



Abdominal image with fat saturation



Abdominal image with fat saturation

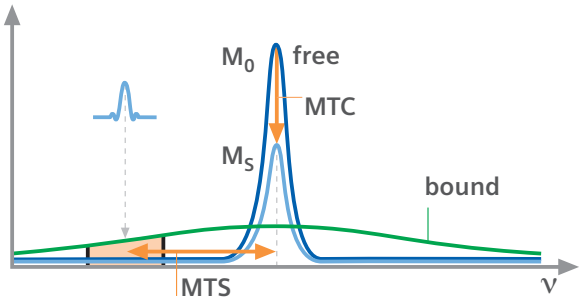


Magnetization transfer contrast

Magnetization transfer contrast (MTC) is an indirect form of saturation. The signal from certain solid tissue, e.g., brain parenchym, is reduced, the signal from liquid components, e.g., blood, is maintained.

Transfer of saturation ...

Protons bound to macro molecules with a very high molecular weight have a wider resonance spectrum than “free” protons. Using a preparation pulse slightly shifted with respect to the resonance frequency, bound protons can be saturated without immediately affecting free protons.



This saturation in itself does not affect the MR image. The bound protons do not contribute significantly to the signal due their large spectral width and low amplitude.

Their special feature is: The saturation is transferred from the bound protons to adjacent free protons (Magnetization Transfer Saturation, MTS). The signal of the free protons is reduced.

... modifies contrast

The signal is reduced in solid tissue due to MTC technology. Blood and fluids are not affected. As a result, contrast between the two components increases. The vessels are more visible.

MTC is used in angiography.

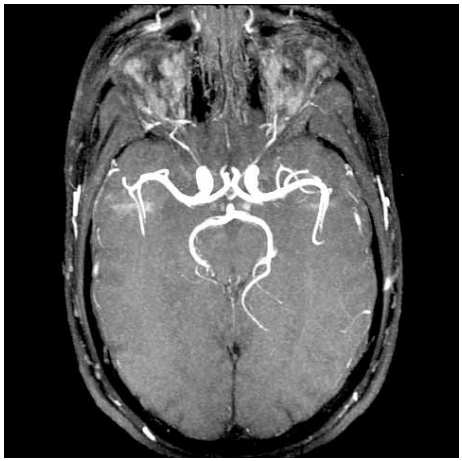


Image without MTC.

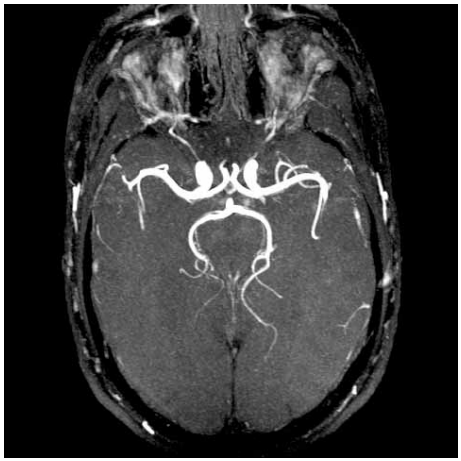
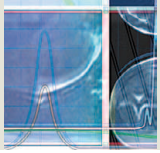
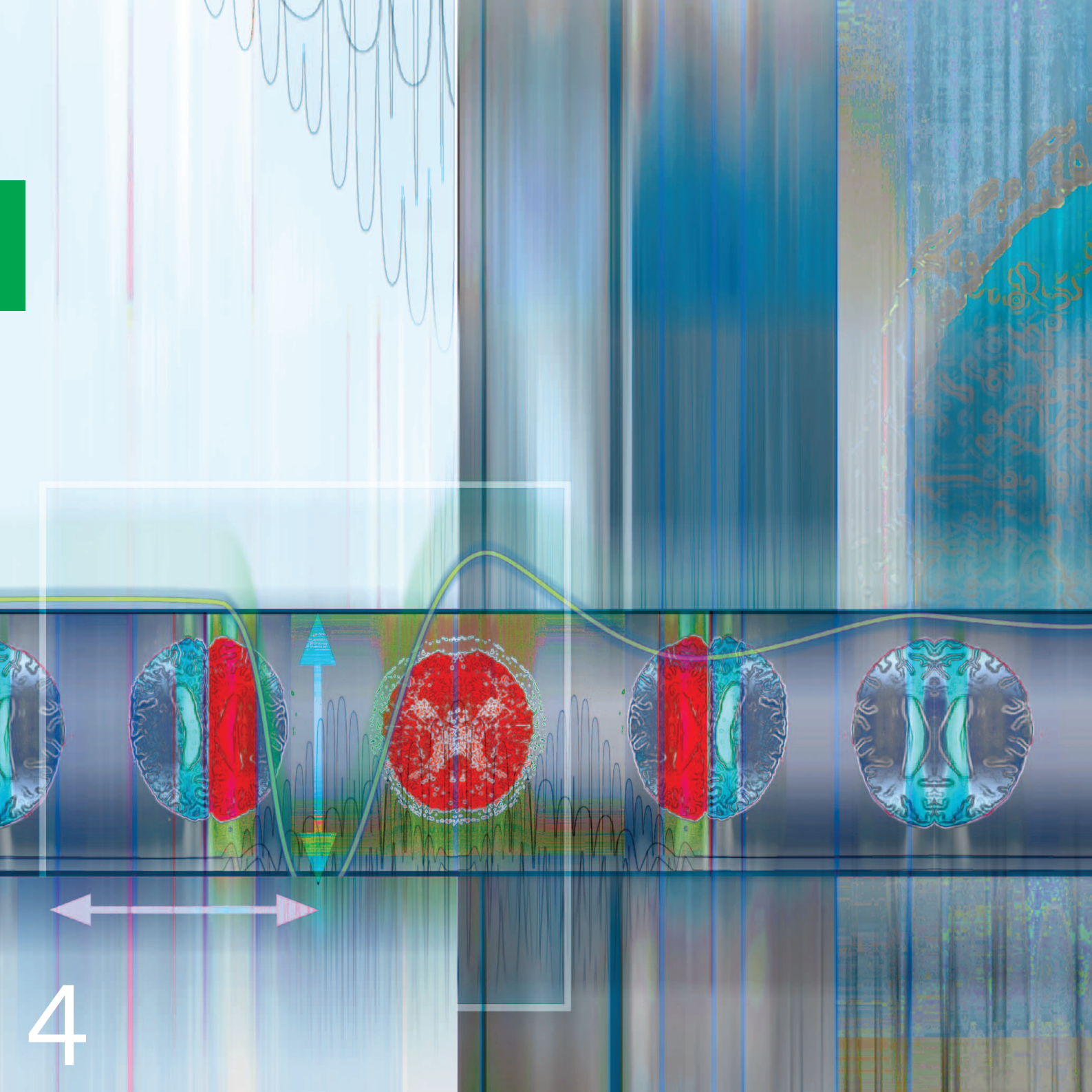


Image with MTC.



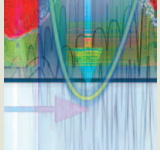
Summary

Saturation can suppress more than motion or flow artifacts. It can also be used for improving contrast in MR images. By controlling saturation accordingly, MR images can display the anatomy or pathology of the slice examined in a more targeted manner.



Functional imaging

Functional imaging—at first this could make you think of the beating heart or the examination of joints. Actually in MR, functional imaging frequently represents functional neuro imaging. The contrasts displayed not only relate to anatomical structures but also to functional activities. Neither fluid nor white/grey matter are shown. Instead you see diffusion, perfusion and neuronal activation.



Diffusion imaging

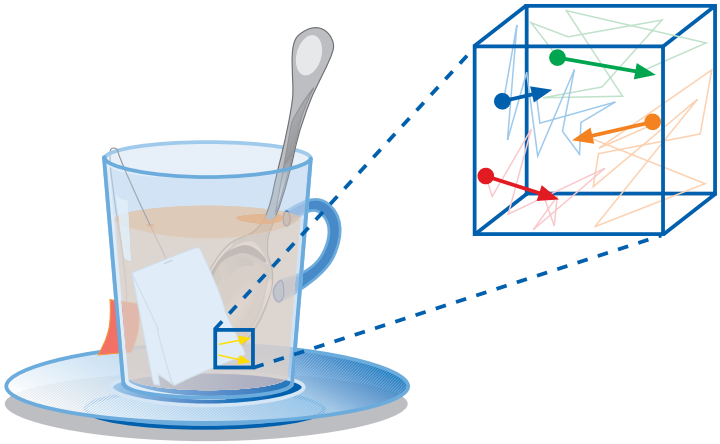
Let's start in the voxel and monitor the molecular motion in the cerebral tissue. There is a law governing the random motion of molecules. Diffusion.

What is diffusion?

DIFFUSION is the process during which molecules of a solution from regions of higher concentration migrate into regions of lower concentration.

You can watch this phenomenon while making yourself a cup of tea. Slowly dip the teabag into a glass of hot water. Although the water was not put into motion, from the color of the tea you can see how it begins to expand more and more in the water.

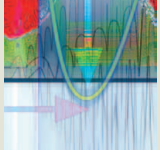
The motor of this molecular migration is known as Brownian motion (thermal random motion).



Incoherence and net shift

The molecules do not move in a straight line. They collide frequently or are deflected. They lack coherent movement.

Diffusion deflects the molecule in the direction of lower concentration. If you monitor molecules for a longer period of time, you will see a linear net shift.



The diffusion coefficient

We are returning from our glass of tea to the human brain. Declines/inclines in concentration exist within tissue, e.g., areas that are high on nutrients versus those that are low on nutrients. These declines/inclines enable molecules to diffuse in a specific direction.

The diffusion coefficient

The average net shift of the molecules depends on the respective tissue. It is stated together with the diffusion coefficient. The DIFFUSION COEFFICIENT is a measure of the mobility of the molecules within certain tissue types.

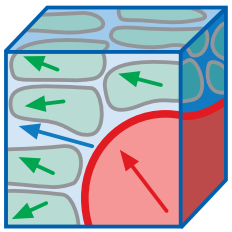
Diffusion coefficient of water in the brain

Liquor	approx. 3×10^{-3} mm^2/s
Gray matter	approx. 0.8×10^{-3} mm^2/s
White matter	as a function of direction $0-1.1\times10^{-3}$ mm^2/s

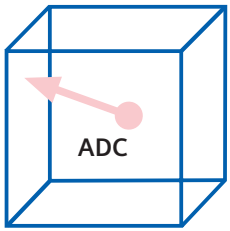
As a function of direction ➔ page 82

Apparent Diffusion Coefficient (ADC)

The diffusion coefficient within a voxel is a mixture comprising intra- and extra-cellular as well as intravascular spin ensembles. These spin ensembles have different diffusion coefficients. Diffusion in a voxel is actually a heterogeneous process.



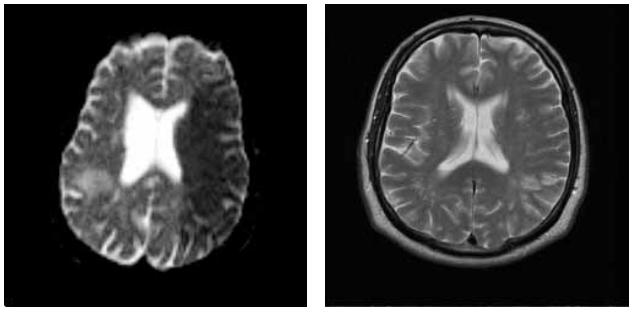
For this reason, the value determined is expresssed as APPARENT DIFFUSION COEFFICIENT, or ADC. It is the *time-averaged* diffusion coefficient of the voxel.



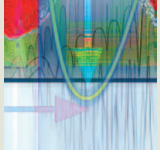
The diffusion image or ADC map

In images, ADC values are shown as the gray values of the pixels. Since these pixels represent the coordinates of the voxels, diffusion displays are similar to anatomical images. However, the signal does not contain any T_2 portions. This is why ADC images are also known as ADC MAPS.

A dark pixel respresents a voxel with low ADC and consequently low diffusion. A bright pixel represents a high ADC and therefore high diffusion.



ADC map with reduced diffusion in the right brain (left) and anatomical T_2 image (right).



The diffusion-weighted MR signal

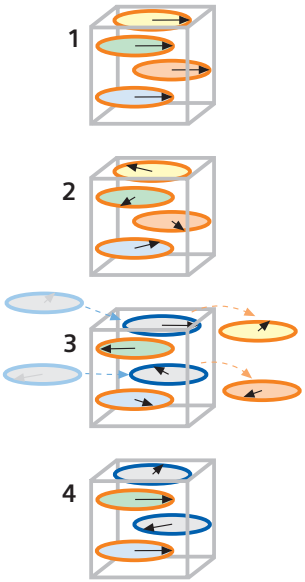
Let's take a look at generating MR diffusion images. The molecular migration is determined with two gradient pulses that are switched briefly, one after the other. The only difference between the two pulses is their algebraic sign, that is, their polarity. Together they form the bipolar diffusion gradient.

How do you generate a diffusion signal?

At first, the spin ensemble of a voxel (1) is completely dephased by a gradient pulse (2). Subsequently, several spins diffuse out of the voxel and are replaced by spins from adjacent voxels (3).

We now apply a gradient pulse with the opposite algebraic sign to the first pulse. Non-migrated spins are completely rephased. Migrated spins of an originally different spin phase cannot be fully rephased by a negative gradient pulse. The signal of the new spin ensemble is reduced (4).

Usually T_2*-weighted sequences are applied to display diffusion contrasts.

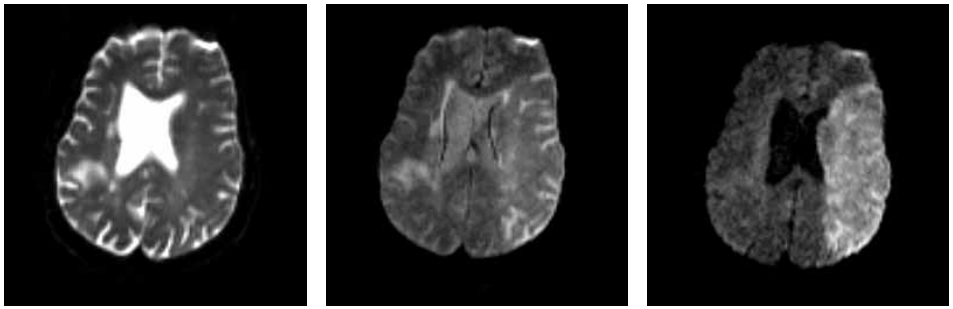


Controlling diffusion contrast

Normally signal attenuation is hard to measure even though diffusion may be strong. For this reason it has to be amplified by the respective parameters of the bipolar gradient pulse:

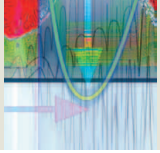
- Strength (Amplitude)
- Duration
- Time interval between the two single pulses

The increase in signal decay is expressed through the value of DIFFUSION WEIGHTING (b-VALUE). An increase leads to a more pronounced signal decay with increasing diffusion.



Diffusion-weighted images: b=0 s/mm^2 (left), b=500 s/mm^2(center), b=1000 s/mm^2(right).

The value b=0 indicates that gradient pulses have not been switched and subsequent diffusion weighting will not be performed (T_2 comparative image).



Diffusion-weighted images versus ADC maps

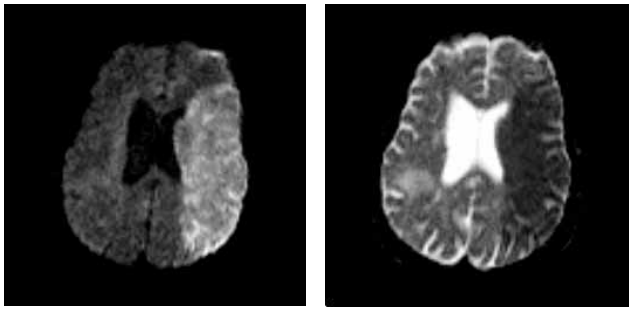
Did you already notice it? The diffusion-weighted image with b=1000 shows diffusing disorder in the right brain. However, this area is darker in the ADC map. Yet both images show the same diffusing disorder.

Diffusion contrast

ADC maps do *not* display anatomical information. Instead they show the diffusion coefficient as functional information. Normal diffusion-weighted images (DW images), however, show anatomical information as well since they contain T_2-portions.

Why does low diffusion appear bright in DW images? The stronger the diffusion, the more spins will be replaced by other voxels. Fewer spins can be completely rephased in the new spin ensemble.

> In DW images, stronger diffusion means a weaker signal. Vice versa, at low diffusion, more rephasable spins mean a higher signal.



Diffusion-weighted image (left) and ADC map (right).

DISCUSSION

Problems with the anatomy

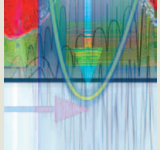
Why aren't we satisfied with anatomical DW images? Diffusion-weighted images contain signal portions coming directly from the tissue independent of diffusion. If tissue has a long T_2- constant, the signal may be increased in the respective region. This increase in signal may be erroneously interpreted as reduced diffusion. This effect is known as *T_2-Shine Through*.

ADC maps are functional images and do not contain anatomical signal portions. This eliminates the possibility of an erroneous diagnosis on the basis of a T_2-Shine Through.

The road from DW images to ADC map

DW images are the basis for computing ADC maps. You begin by comparing two different diffusion-weighted images and compute the fit across a theoretical exponential curve. To eliminate the perfusion flow rate, DW images are selected that show a b-value above 150 s/mm^2. In this range, perfusion spins show a complete signal loss due to dephasing. In addition to the reference image (b = 0 s/mm^2) DW images with b-values of 500 mm/s^2 and 1000 mm/s^2 are generated.

However, there are additional complications, noteably the anisotropy of diffusion→ page 82.

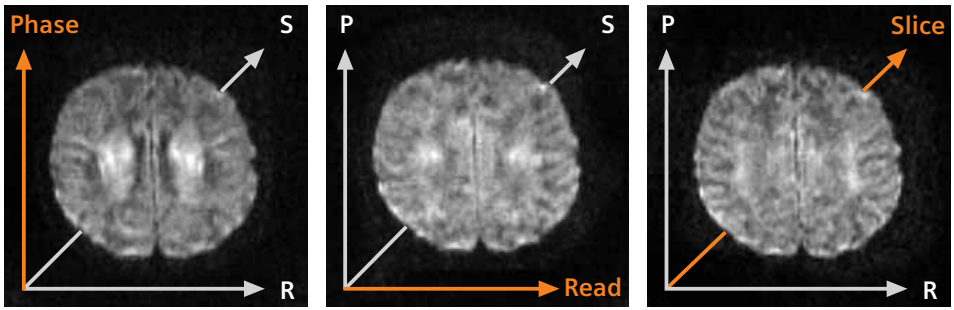


Diffusion and gradient orientation

In tissue, diffusion of water molecules is not always possible. For examples, cell layers adjacent to tissue boundaries may limit diffusion. The signal intensity depends therefore on the orientation of the diffusion gradient.

Anisotropy

Myelin provides an example for diffusion as a function of direction. The myelin sheath surrounds the myelinated nerve fibers. Only very few water molecules can pass through the sheath. For this reason, diffusion is largely limited *in the transverse orientation* to the fibers. There are none or very few limitations *in the longitudinal direction* of the fibers. Diffusion is therefore anisotropic, that is, spatially disparate.



Alignment of diffusion gradient in the phase-encoding direction (phase, left), read-out direction (read, center) and slice selection direction (slice, right). All three images show the same slice.

This becomes visible by switching the diffusion gradients into the three orthogonal directions. Due to the anisotropy, the same slice shows a different diffusion contrast as a function of the spatial direction selected.

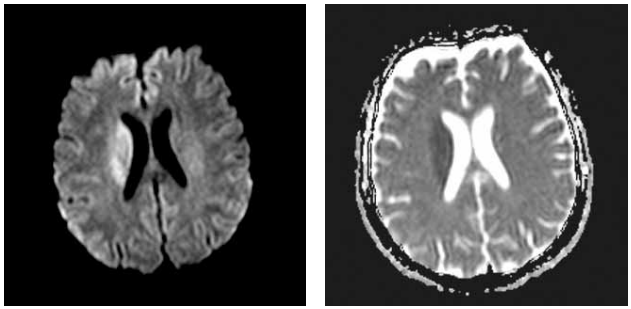
Averaged diffusion

To display diffusion, we frequently need images independent of anisotropy.

The TRACE IMAGE is the average of the three measurements in orthogonal directions. This is the easiest way of generating diffusion images that are independent of anisotropy.

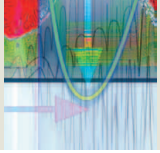
An averaged ADC map is generated the same way. The average of the three ADC maps in the orthogonal direction results in an averaged ADC map.

It is important to determine the trace value, especially for follow-up examinations. A slight change in patient positioning will lead to an offset alignment of the tissue structure as compared to the diffusion axes.



Trace image (left) and averaged ADC map (right).

Due to its higher diagnostic value, the averaged ADC map has almost fully displaced the standard ADC map.



DISCUSSION

The diffusion tensor

An anisotropic magnitude is mathematically expressed as a tensor. This is so-to-speak the next dimensional leap after scales and vectors. As a result, the diffusion tensor is usually shown as a square matrix that consists of nine numbers (3x3 matrix).
The three diagonal elements of the matrix represent diffusion in the three orthogonal directions. The sum of these three diagonal elements (Tensor Trace) result in the well-known trace image.

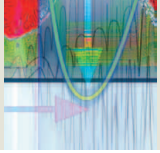
The matrix elements to the right and left of the diagonals are populated by values that only differ with respect to their algebraic signs. Under normal circumstanes, you only need diffusion gradients in 6 directions to fill the entire diffusion tensor. For displaying complicated anisotropies, you'll need to apply as much as 12 or more different diffusion gradients.

This technique is known as Diffusion Tensor Imaging (DTI) or Multi-Directional Diffusion Weighting (MDDW). With the help of this technique, you may delimit or selectively display the fiber connections of white matter or the individual core areas of deep grey matter.

Reduced to the essentials

Diffusion is shown in MR imaging through diffusion-weighted images (DW-images) and/or the averaged diffusion coefficient ADC. Since diffusion may also include a component dependent on direction, an additional averaging is performed across the orthogonal directions. The results are trace images and/or averaged ADC maps.

ADC maps do not show anatomical but rather functional information.



Perfusion imaging

From the intracellular perspective of diffusion we are now moving to the higher plateau of perfusion. For this purpose, we are using a gadolinium-based contrast agent to visualize the processes.

Contrast agent, First Pass and signal

PERFUSION is the vascular transport of nutrients for supplying the cells in the capillary bed of tissue.

Perfusion is traced by intravenously injecting a contrast agent (CA). The CA bolus reaches the brain within a short period of time. While it passes through the cerebral capillary bed (FIRST PASS) the perfusion of the bolus is visible. In case of satisfactory perfusion, the contrast agent is finely distributed in tissue via the capillaries and is washed out by the blood that follows.

During first pass, the signal does *not* increase as may be expected. Instead the *signal decreases* significantly with T_2*-sequences.

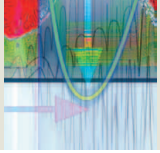
DISCUSSION

Susceptibility contrast

The reason for the reduced signal is the change in relaxation rate R_2*, the reciprocal value of T_2*. This change in R_2* is the result of the different magnetization of the intravascular space filled with contrast agent and the surrounding tissue, the differences in susceptibility of both areas. This is why this imaging method is also known as Dynamic Susceptibility Contrast Imaging (DSC).

Perfusion in MR

Biologically seen, perfusion is the flow of nutrients in the capillaries. The flow begins with the vascular transport (from the arterioles) into the capillary bed. It is followed by a diffusion of the molecules of the nutrients through the cell membranes into the cell to be supplied. The reperfusion transports the waste product from the cells to the lympathic system via the capillary bed.
Within the context of MR imaging, perfusion describes only the vascular transport phase. It does not describe the diffusion phase of the perfusion.

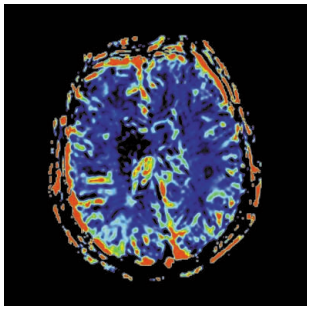


Blood volume and blood flow

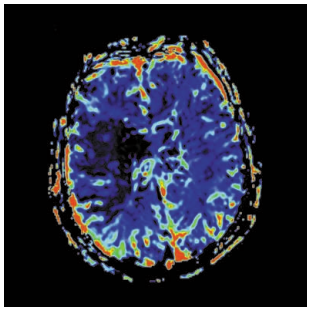
The regional blood volume and blood flow in the brain are the two main parameters for perfusion imaging.

The REGIONAL CEREBRAL BLOOD VOLUME (rCBV) is the space occupied by the capillary bed within a voxel. It refers to the dimensions of the tissue supplied (measurement unit: ml/g).

The REGIONAL CEREBRAL BLOOD FLOW (rCBF) represents the amount of blood flowing within a specific period of time through the capillary bed within the voxel (in ml/g/s).



rCBV map.

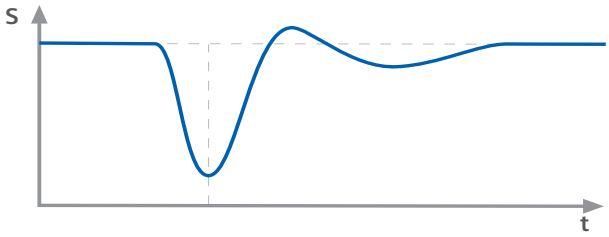


rCBF map.

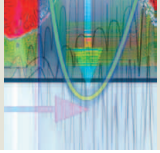
Global bolus plot

The progression over time of the First Pass as well as the associated signal reduction are displayed for the entire brain (not for the individual voxel) as a time-density curve, the GLOBAL BOLUS PLOT (GBP).

As a first step, GBP evaluates both bolus passage and perfusion. During the second step of the evaluation, local differences are examined.



Global Bolus Plot (GBP).



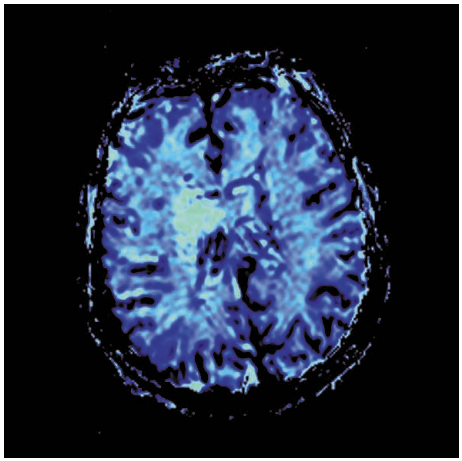
Perfusion images

The Global Bolus Plot provides only a general statement with respect to the course of perfusion over time. Individual voxels have to be evaluated first to provide precise data with respect to cerebral blood volume and blood flow. Individual cards and maps are generated for each slice to be included in the evaluation. The two most important maps are shown below.

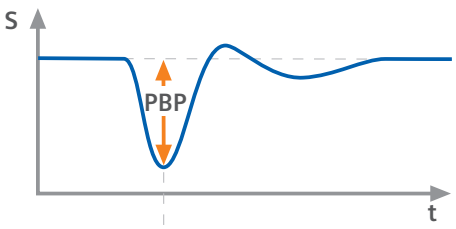
Reduced perfusion

The PERCENTAGE OF BASELINE AT PEAK (PBP) determines the *relative* amount of signal loss due to bolus passage through the capillary bed. One PBP map per slice measured is shown.

Reduced perfusion, that is, less contrast agent, is shown as bright pixels.



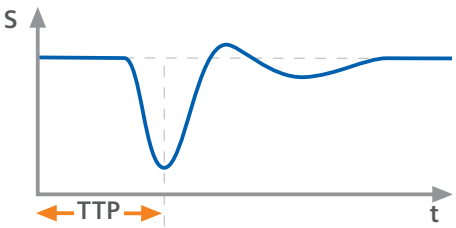
PBP map.

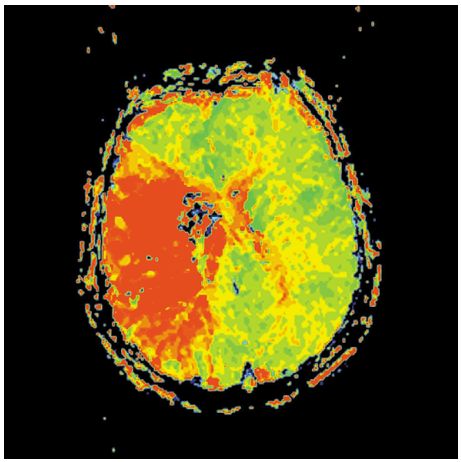


Delayed perfusion

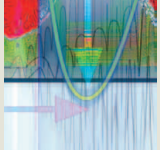
The TIME TO PEAK (TTP) is the time interval to the bolus peak. In the TTP map, the signal intensity of each pixel shows in grey or in color-encoding the regional distribution of the time from the injection of the contrast agent to the bolus peak.

Brighter pixels (for gray value encoding) represent a delayed TTP and therefore delayed perfusion.





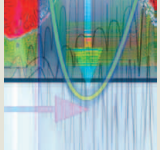
TTP map in color, the red pixels indicate delayed perfusion.



Reduced to the essentials

The First Pass of a contrast agent through the cerebral capillary bed visualizes perfusion. Regional blood volume as well as regional blood flow play an important role.

Perfusion is evaluated with the PBP map and the TTP map. They do not show anatomical but rather functional contrast.



BOLD imaging

What are you thinking right now? You are reading and you are processing stimuli that the retina is forwarding to the brain. What regions of the brain are processing these senses? This question can be answered with BOLD imaging.

Blood is becoming a signal carrier

Which neurons are participating in the reading process? When searching for the respective brain region, we are *not* measuring the activity of neuron assemblies directly, instead we are looking for locally increased oxygen concentrations connected with changes in blood circulation (BOLD = Blood Oxygenation Level Depending).

We are also *not* measuring oxygen consumption. *After* having passed the neurons, only the oxygen contained in the blood determines the signal.

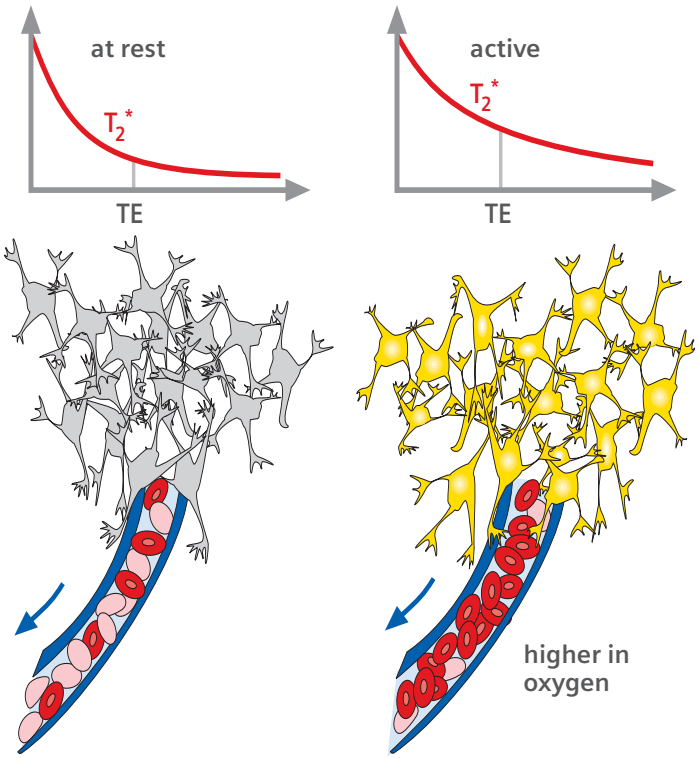
During activity, the cerebral neurons' need for oxygen increases. The increase in blood flow connected with this ensures that the blood contains more oxyhemoglobin, the carrier of oxygen, after oxygen depletion than during the inactivity of neurons.

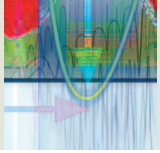
Oxygen depletion is overcompensated. An inrease in oxygen concentration in a brain region is therefore an indicator for active neuron assemblies.

Susceptibility changes in blood

Through oxygen enrichment, the magnetic characteristics of the blood change: Blood low in oxygen contains more paramagnetic desoxyhemoglobin (HB^{++}), blood high in oxyen contains more diamagnetic oxyhemoglobin (HBO_2^{--}).

With increasing oxygen, the magnetic characteristics of hemoglobin adjust to the surrounding blood plasma. As a result, transverse magnetization decay slows down: T_2* is extended, and the signal increases.





Paradigm, t-test images and mosaic images

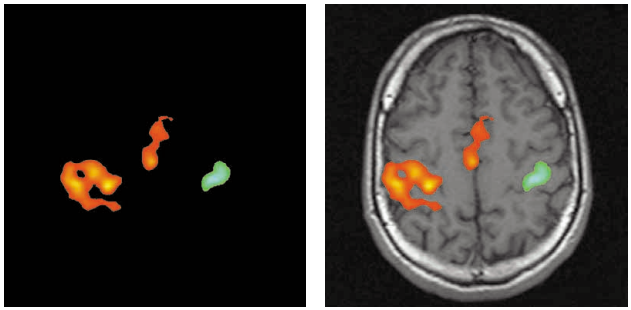
When we talk about an increase in oxgen during activation, we also have to know the oxygenation level that can be measured in the individual brain regions without activation, that is, at rest. Without this comparison, we canot make a statement.

From signal to image

Let's assume, we would like to determine the brain region that is active while we are moving our fingers. During the measurement we generate acquisitions for several seconds *without* moving any of the fingers. We follow that with acquisitions were the fingers are being moved for several seconds. To obtain valid results, we perform this interchange several times, e.g., 10 times—that is to say we perform a PARADIGM.

The images taken at rest and those with neuronal activity are subtracted with the t-test, a statistical method. The images computed are purely functional and not anatomical.

To be able to allocate the signal to the respective brain region, the t-test image is usually superimposed on the anatomical image.



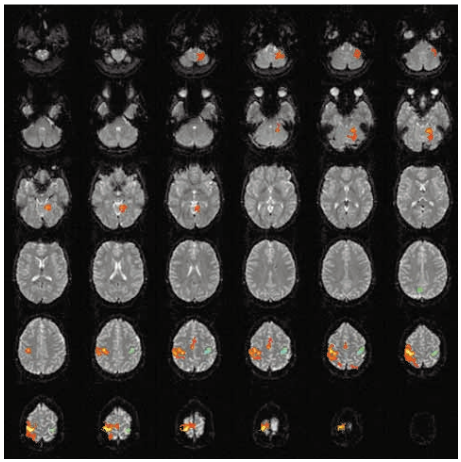
t-test image (left) and superimposed image (right).

Even the slightest motion of the head may already falsify measurement results. By using the three-dimensional motion correction 3-D PACE (Prospective Acquisition CorrEction), the offset in the images caused by motion is already corrected during the measurement (inline technology).

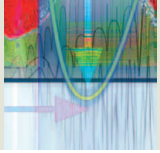
From image to mosaic

To detect active brain regions, images of the entire brain are required. Each measurement generates a multitude of images from a series of slices. A total of thousands of images are measured and computed.

To be able to manage this volume on images, mosaic imaging is implemented. The software combines and stores the images into a matrix within the measurement sequence.



Mosaic imaging.

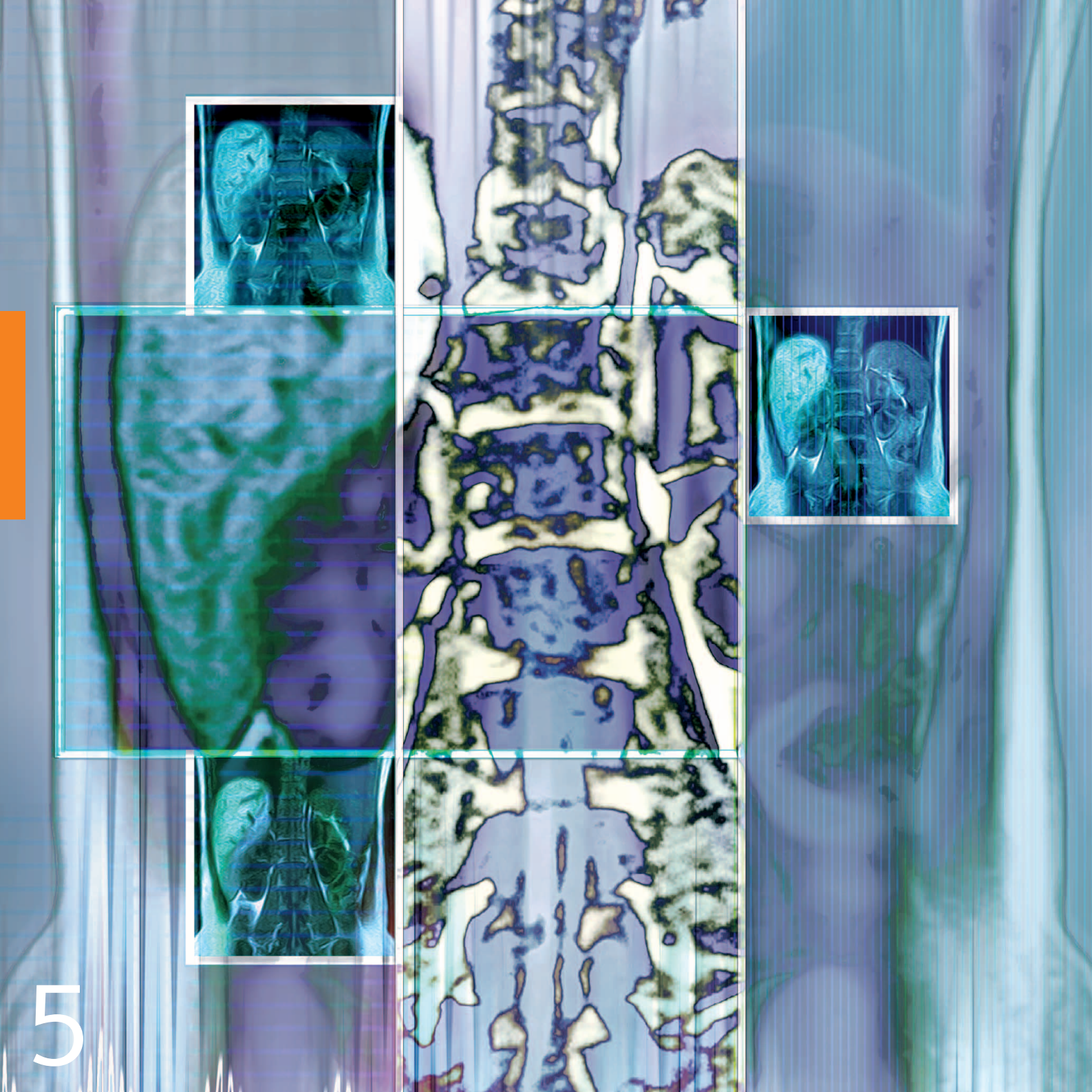


Summary

Neuron assemblies, blood flow and nutrient transport, as well as molecular motion — functional imaging provides us with new insights into the human brain.

Research involving epilepsy, evaluation of vessel closures or the diagnoses of strokes are a number of applications addressed by this branch of MR imaging.

Ultra-high field systems with a field strength of 3 Tesla and higher are able to improve visualization of processes on the molecular and neuronal level. This leads us to expect that the importance of functional imaging is increasing steadily.

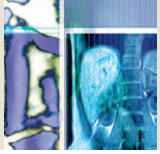


Fast imaging techniques

In the chapter covering image quality, we have seen that compromise involving measurement time, resolution and SNR provide for an optimal image.

In addition, the sequences used include different features.

You are viewing the results of several selected sequences and their applications.



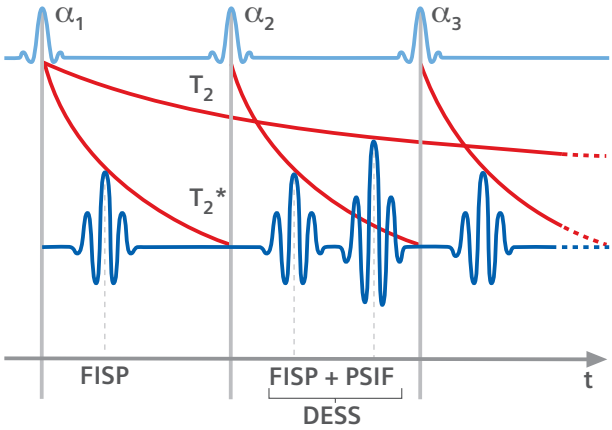
Variants of the gradient echo technique

In *Magnets, Spins and Resonances* we have seen that the gradient echo technique is basically faster than the spin echo technique. What happens when we combine measurements or echoes?
As it applies to all gradient echo sequences, the following types of sequences are available as 2-D as well as 3-D applications.

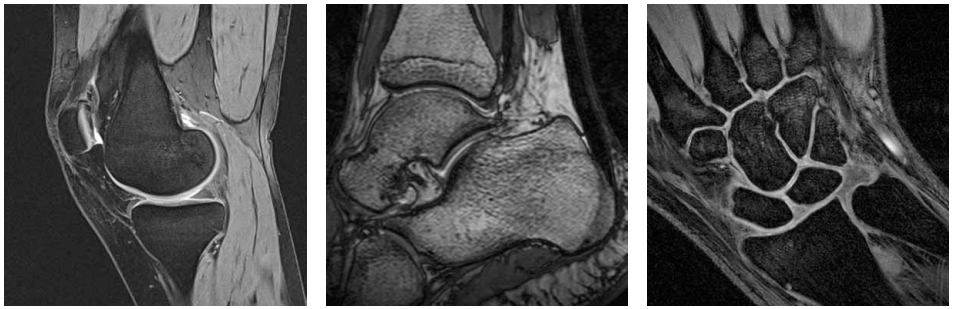
Combining two gradient echoes (DESS) ...

Dual **E**cho **S**teady **S**tate: The DESS sequence generates two echoes in the same repetition time, one FISP and one PSIF echo.

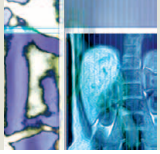
FISP is T_1/T_2*-weighted; PSIF is strongly T_2-weighted. By combining both echoes, the DESS sequence obtains a better T_2 contrast than a pure FISP sequence.



... enhances contrast



DESS provides two advantages: We are measuring two images, improving the SNR. And the combination FISP and PSIF provides for a strong T_2 contrast with T_1/T_2* weighting. This allows for good differentiation of synovial fluid and cartilage — an important element in orthopedics.

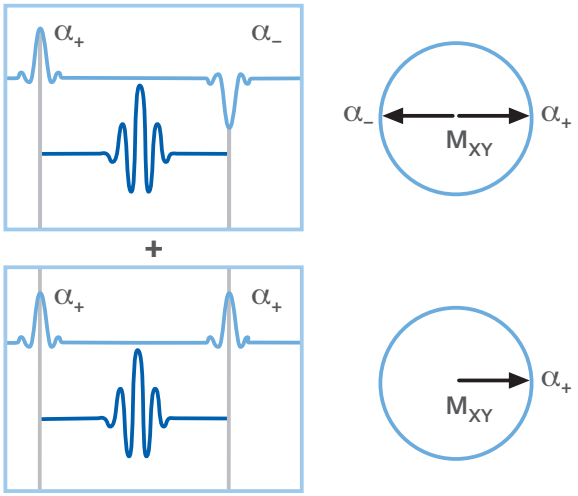


Variants of the gradient echo technique: CISS

Constructive **I**nterference in the **S**teady **S**tate: The CISS sequence provides images at submillimeter resolution with strong T_1/T_2 contrast.

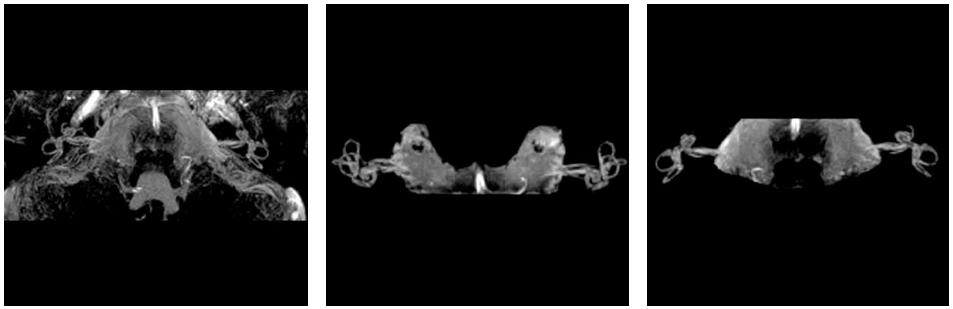
Constructive interference ...

CISS combines two TrueFISP sequences. TrueFISP provides the highest signal of all steady-state sequences. During one measurement the phase angles of the RF pulse are alternated (α_+, α_-), in the other measurement they are *not* alternated (α_+, α_+). This is how images with two different echo types are generated and subsequently superimposed. The sum images are T_1/T_2-weighted and may include interference stripes. By combining two measurements in CISS, these stripes are eliminated from the images.

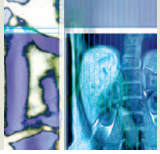


... avoids stripes in images

CISS-3D offers: Submillimeter-resolution with a very high SNR for fluids. The sequence is robust with strong T_1/T_2 contrast. Typical applications include the inner ear and the cerebellum.



This demonstrates one of the advantages of the 3-D technique over the 2-D technique: With thin slices, the higher SNR allows for excellent detail detection of anatomical structures.



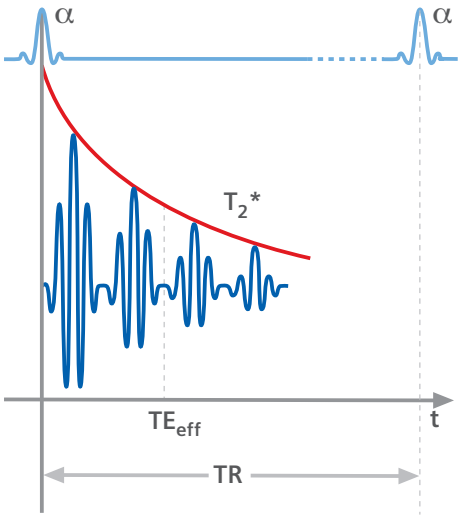
Variants of the gradient echo technique: MEDIC

Multi **E**cho **D**ata **I**mage **C**ombination: MEDIC supplies T_2* contrast at high resolution. Flow artifacts and effects caused through chemical shift are reduced.

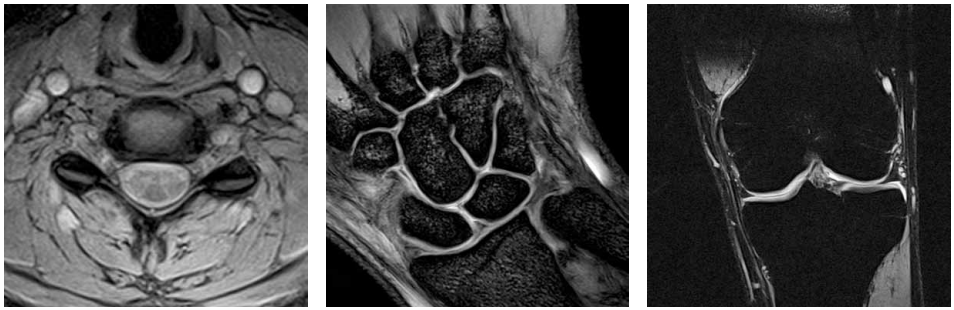
Generating several images ...

MEDIC is a multi-echo sequence. Flow effects are compensated with each echo. The sequence combines several images with different T_2* contrast.

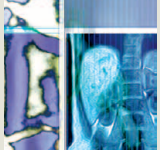
Since images with different echo times are added up, the new images show a mixed T_2* contrast. For this reason, only an effective echo time TE_{eff} can be provided as echo time.



... with different T_2* contrast.



MEDIC does not only minimize flow artifacts. The sequence also reduces artifacts connected with the chemical shift.
It provides T_2* contrast at high resolution. Typical applications involve the cervical spine and orthopedic requirements.



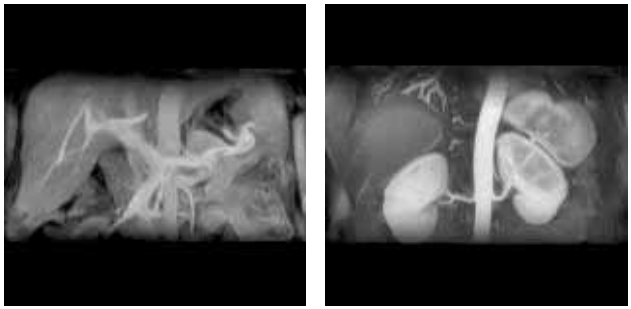
Variants of the gradient echo technique: VIBE

Volumetric **I**nterpolated **B**reathhold **E**xamination: The T_1-weighted VIBE sequence uses breathhold technique for avoiding artifacts. This 3D-Flash sequence obtains high spatial resolution.

High image quality despite short measurement time ...

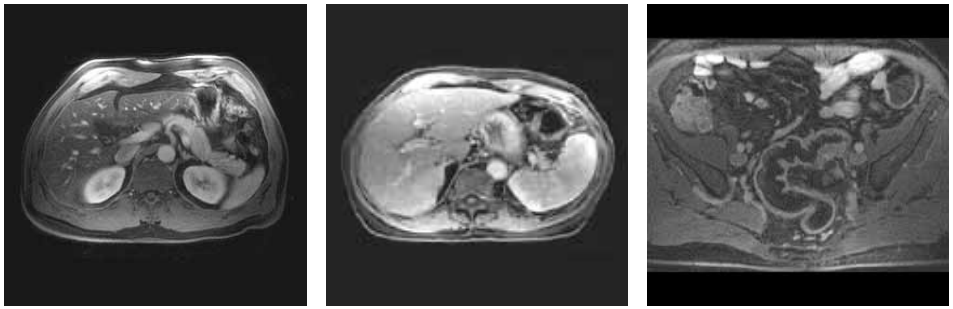
VIBE combines two techniques: Half-Fourier technique accelerates the measurement of partitions. The three-dimensional measurement is accelerated by interpolating the measurement points in the slice selection direction.

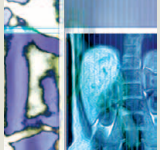
Although GMR flow compensation ➔ page 41 is omitted, an angiographic image effect is obtained based on the extremely short echo times. VIBE always uses fat saturation. With dynamic contrast agent studies, the fast VIBE sequence provides for timely precision in the acquisition of vessels in the arterial and venous phase, especially in the abdominal area.



... for abdominal imaging

The VIBE sequence facilitates artifact-free acquisition in breathhold technique of the abdominal region. Additional areas of application are the thoracic and pelvic regions.





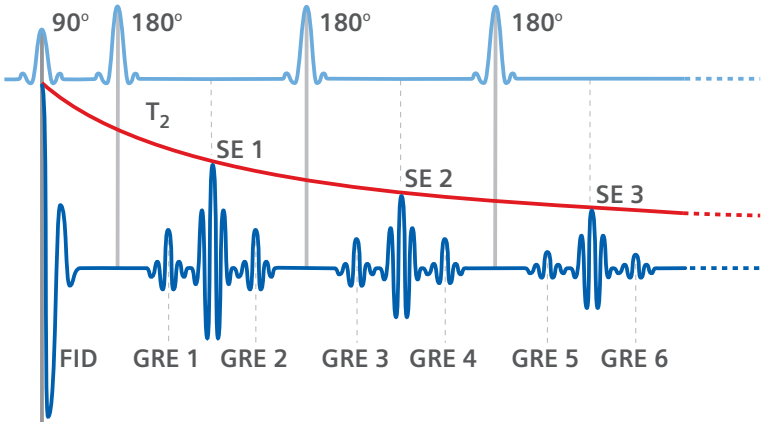
Turbo techniques

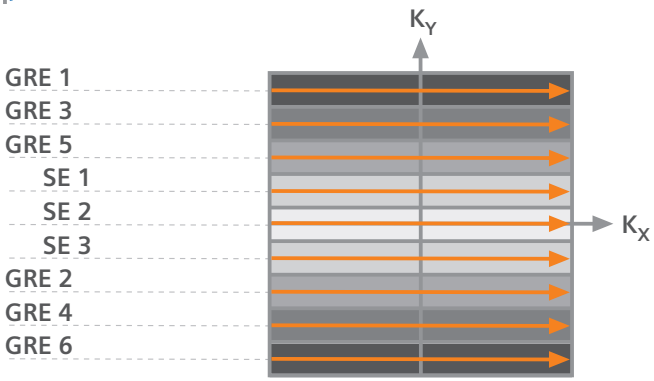
The basic principle of turbo techniques: A single stimulation measures several echoes. We've explained this in *Magnets, Spins and Resonances*.

Using additional gradient echoes (TurboGSE) ...

TurboGSE is an expansion of the turbo spin echo technique. A TurboGSE sequence generates additional gradient echoes before and after each spin echo, using dephased and rephased gradient pulses switched accordingly.

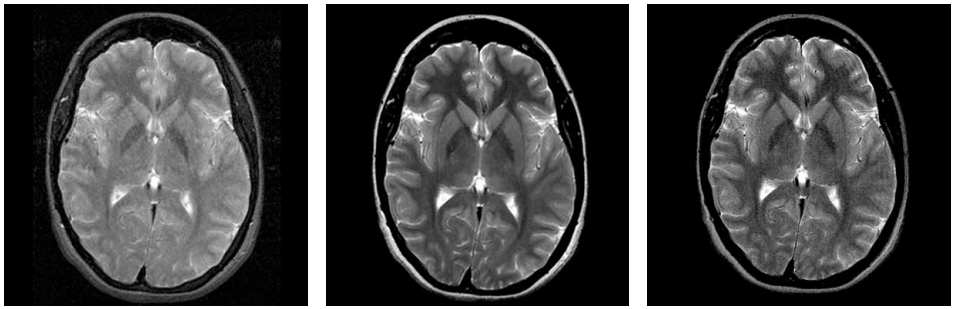
How are the spin echoes and gradient echoes filled into the k-space?
The spin echoes provide the center segments and ensure contrast. The gradient echoes determine the resolution of the outer segments.



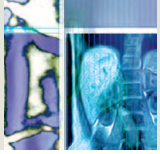


... Reducing the measurement time even further

The additional echoes allow us to measure faster. Or we measure more slices within the same time period. There is no amplification of the fat signal as is the case with TurboSE-technique.



Here are the results of T_2 spin echo (left, measurement time of 7 inutes), T_2 TurboSE (center, measurement time of 8 seconds) and T_2 TurboGSE (right, measurement time of 6 seconds).



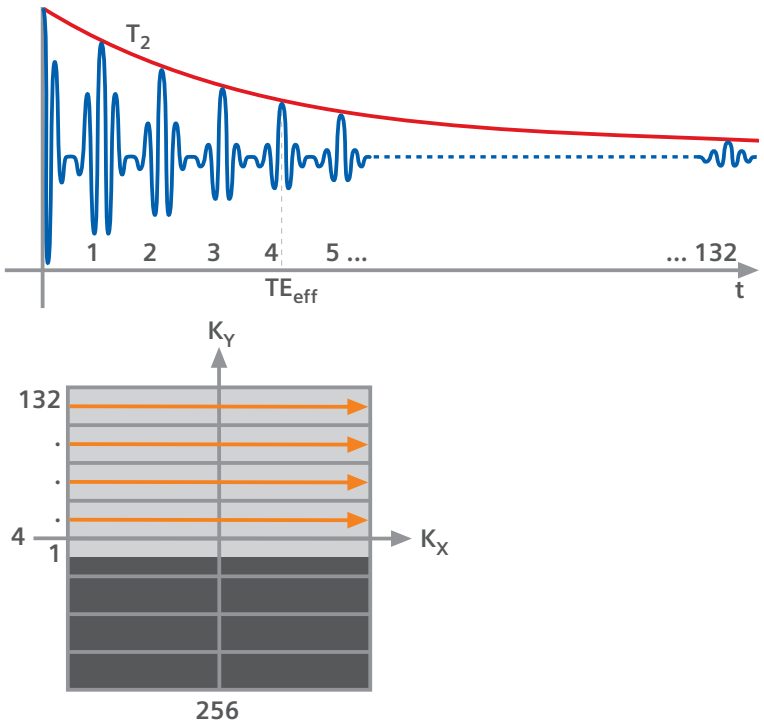
Turbo techniques: HASTE

Half **A**cquisition **S**ingle shot **T**urbo spin **E**cho:
Fast measurement, high resolution and T_2 contrast. All of that is supplied by HASTE.

Using nothing more than a single excitation ...

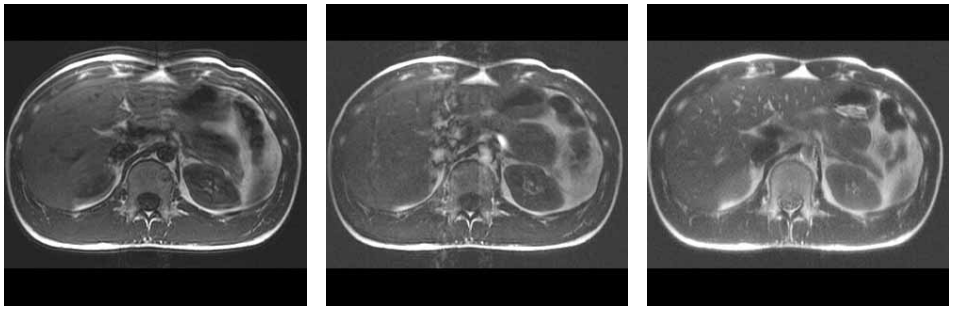
HASTE combines two techniques. This TurboSE-sequence uses a single excitation to generate all echoes for an image (single shot). The first echoes are encoded via small phase-encoding gradients slightly below and above the center raw data line; the gradient strength is increased from echo to echo until the upper half of the k-space is filled. The contrast is determined by the effective echo time TE_{eff}, that is, the echo time in the center of the k-space.

In addition, the Half-Fourier technique accelerates the measurement further: Only slightly more than half of the raw data are acquired.



... Freezing movements

HASTE reduces the measurement time of a single slice to below 1 second.
The short measurement time keeps artifacts caused by involuntary patient movement or respiratory motion to a minimum.

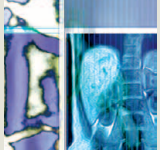


Here are the results of spin echo (left), TurboSE (center) and HASTE (right).

HASTE is suitable for abdominal examinations, for restless patients, or for pediatric patients.

Typical measurement times per slice

SE	3 – 4 minutes
TSE	20 seconds – 2 minutes
HASTE	0.6 seconds



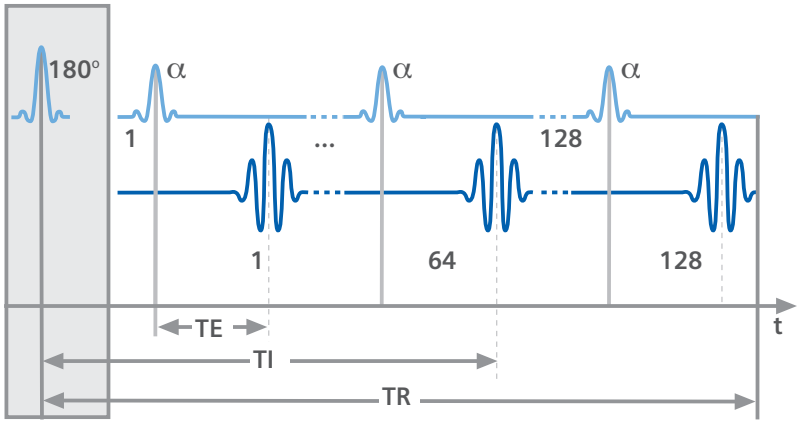
Turbo techniques: TurboFLASH

TurboFLASH reduces motion artifacts by using short measurement times.
The technique allows for dynamic perfusion series after injecting contrast agent and imaging in CINE technique.

Two phases

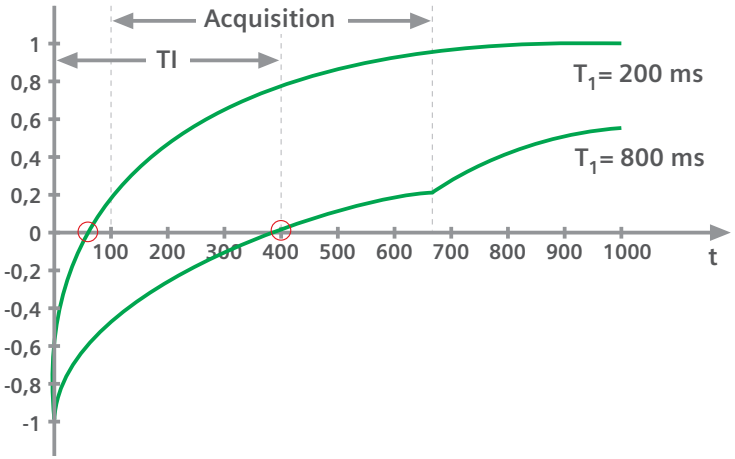
The preparation phase determines image contrast. For example, a 180° inversion pulse is used prior to the actual sequence.

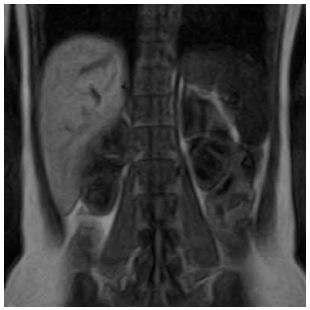
During the acquisition phase, the matrix is measured in one step with a very fast gradient echo sequence.



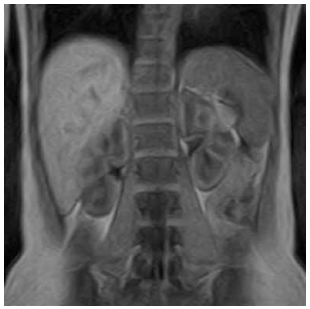
The inversion time determines contrast

The signal from different tissue types can be suppressed with the well-known T_1 constants. Select a suitable inversion time TI (in this case: 400 ms).

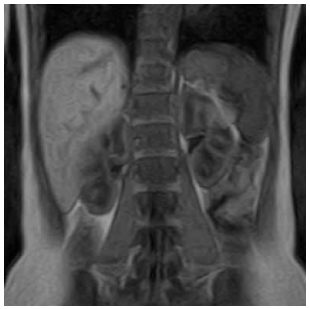




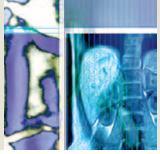
TI = 50 ms



TI = 400 ms

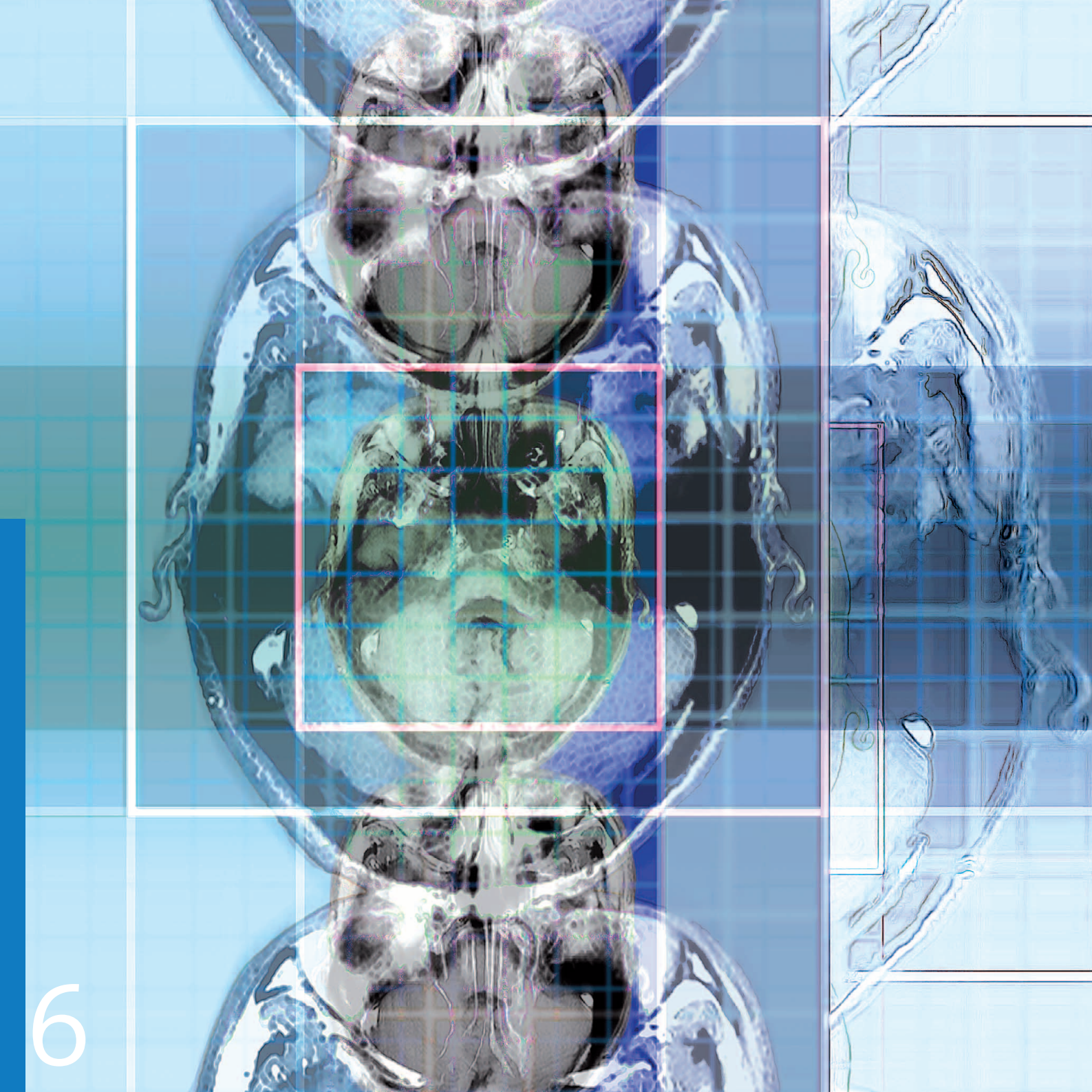


TI = 800 ms



Summary

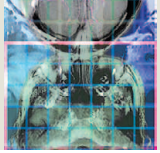
Today MR imaging offers suitable sequences for a multitude of special applications. Important objectives that need to be optimized are: Speed, resolution and image quality. Gradient echo and Turbo techniques can be varied according to the diagnostic question at hand and subsequently cover a wide field of applications.



Detecting and avoiding artifacts

Artifacts are structures in the image that do not correspond to the spatial distribution of tissue in the image plane. To avoid diagnostic misinterpretations, it is recommended to learn how to detect and allocate these artifacts.

Complex MR imaging is familiar with a host of artifact types. They result from physiological, physical as well as system-related influences.



Motion and artifacts

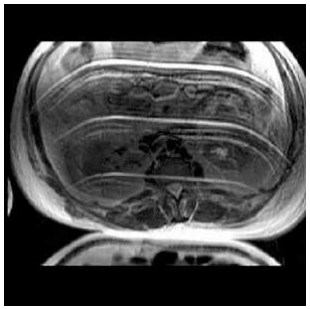
The most pronounced artifacts in the image relate to motion during acquisition: Respiration, heart beat, blood flow, eye movements, swallowing, accidental patient movement.

Ghosting and smearing

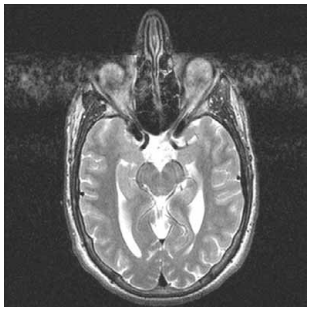
The rise and fall of the thorax during breathing produces GHOSTING: The thorax is seen as a locally offset double or multiple structure. Structures rich in signal, for example, subcutaneous fat, further amplify ghosting.

The aperiodic movements of the eyes continuously add SMEARING to the image.

These motion artifacts are seen exclusively in the phase-encoding direction.



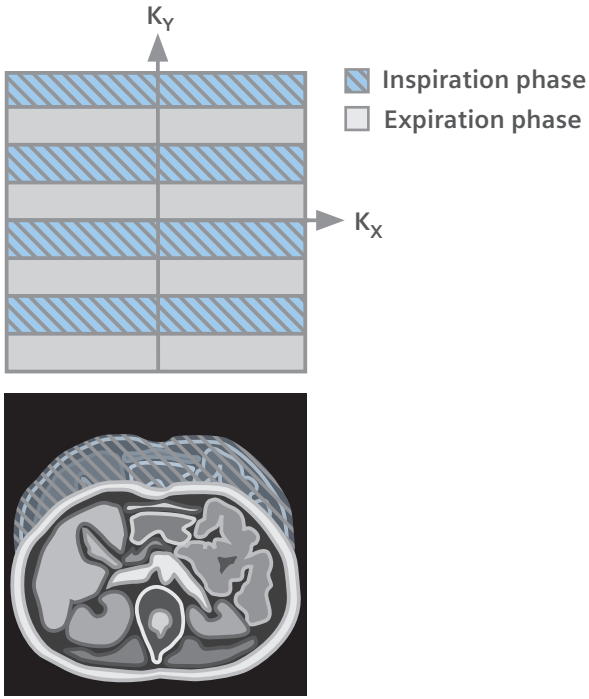
Ghosting in the vicinity of the thorax caused by periodic respiratory movement.

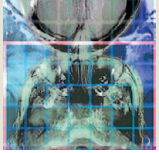


Smearing caused by aperiodic eye movement.

The source is incorrect encoding

During periodic movements such as breathing, the thorax is in several, equally-distanced phase-encoding steps during the inspiration phase. During the steps in between, the thorax is in the expiration phase. This leads to quasi-periodic incorrect encoding: the thorax appears as locally offset in the MR image.



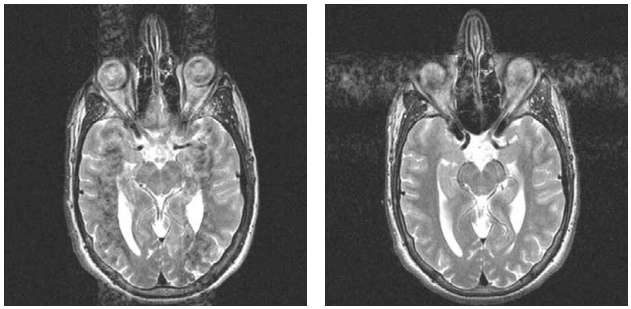


How to remedy motion artifacts?

By swapping phase and frequency-encoding (Swap) motion artifacts can be frequently relocated to image areas that do not affect interpretation.

Other possibilities are:

- Fat suppression
- Image averaging, e.g., LOTA
- Breathhold techniques



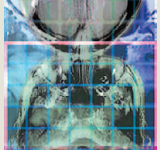
Relocation of motion artifacts by swapping phase and frequency-encoding.

Reduced to the essentials

Motion artifacts can be divided into two groups: Ghosting and smearing.

Ghosting is the result of quasi-periodic motion (example: respiration).

Smearing represents aperiodic structures (example: eye movement).



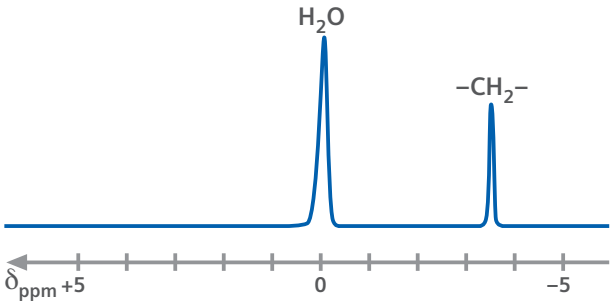
Physically-caused artifacts

Let's take a look at physically-caused MR artifacts. The image shows relief and contour formations as well as distortions. The starting points are the chemical shift and magnetizability (susceptibility). The well-known chemical shift is a good starting point.

Chemical shift

In almost all bio-molecules, several hydrogen atoms are bound to different positions. Different positions mean different chemical and therefore usually different magnetic environments. The local magnetic field is either reduced or increased, the resonance frequencies of the bound protons are slightly lower or higher than the typical Larmor frequency.
This is why the nuclei of a molecule are able to supply *several* resonance lines.

This shift in resonance frequencies is known as CHEMICAL SHIFT. The chemical shift can be recognized by a shift of the allocated resonance lines in the measured spectrum.



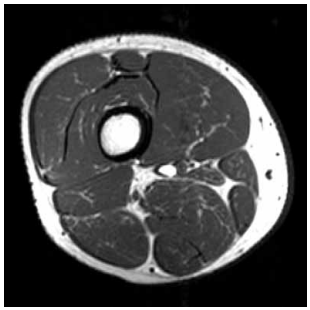
Chemical shift of 3.4 ppm for water and methyl group ($-CH_2-$), the main component of fat. The unit used for a chemical shift is δ_{ppm} (ppm = parts per million). $\delta_{ppm} = -3.4$ means that the frequency of the methyl group is reduced by 3.4 millionth.
In spectral displays common to MR, the frequency axis is oriented from right to left.

Relief artifacts caused by the chemical shift

The relief artifact is caused by the distance between the resonance lines of fat and water protons in the spectrum.

Relief artifacts

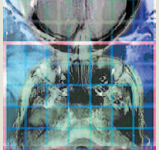
Especially susceptible are tissues with direct transitions between fat and water, e.g., vertebras and intervertrebral disks, as well as transitions between spleen and kidneys and surrounding fat. The artifact is visible in the image as a spatial shift. Since water as well as fat protons contribute to image generation, the artifact is caused by their chemical shift of 3.4 ppm.



Relief artifact.

The source for relief artifacts

Because of their chemical shift, the signals of the fat and water protons in a voxel are allocated to different image pixels during image reconstruction. At transitions of fat and water, these incorrect encodings lead to a higher signal (dark surface area) or to an invalid signal (bright areas) in the respective frequency-encoding direction.



What to do with relief artifacts?

When imaging the brain, the chemical shift is not of import, since the signal intensity of fat is considerably lower than that of water. In all other cases, the following remedies may be used:

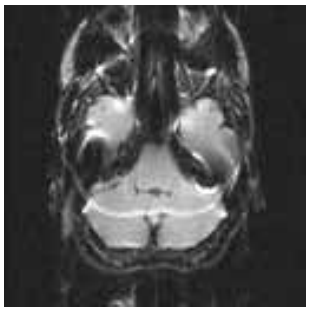
- Sequences with larger bandwidths
- Swapping of phase and frequency-encoding direction
- Use of STIR sequence
- Fat or water suppression

Relief artifacts during EPI imaging

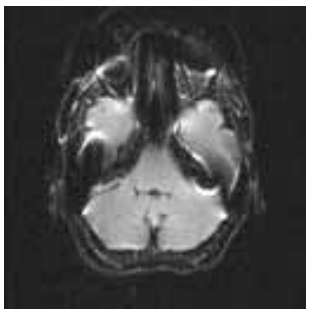
Relief artifacts also occur with echoplanar imaging. Since the fast T_2* decay of the FID (Free Induction Decay) leaves only 100 ms for generating echoes, readout is usually limited to between 64 and 128 echoes.

For EPI sequences with their low bandwidths in the phase-encoding direction, the chemical shift of fat and water is seen as a clearly defined artifact **in this direction**.

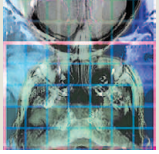
Relief artifacts with 1 Tesla systems	
Readout bandwidth	780 Hz
Phase bandwidth	7 Hz
Shift	21 Pixel



Shift of fat and water in the head.



Artifact elimination in EPI imaging through fat suppression.



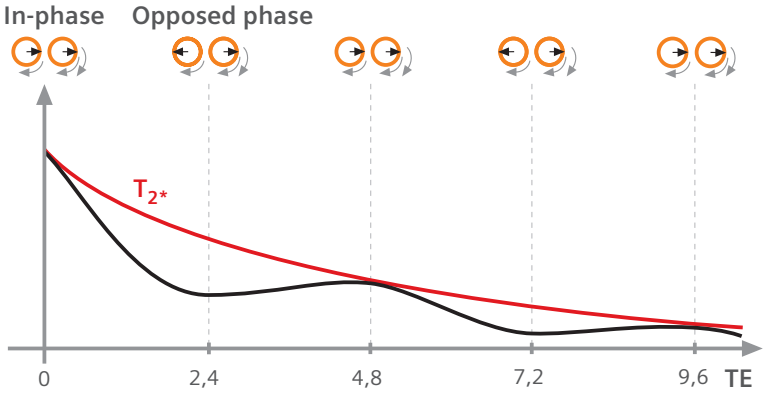
Contour artifacts caused by chemical shift

The chemical shift between fat and water protons is the basis for additional image phenomena. In connection with gradient echo sequences, “phase nulling“ may occur in the image.

Phase cycling

The chemical shift of water and fat protons is the source of a possible phase shift affecting signal generation of a voxel containing fat and water.

With a spin echo sequence, the protons in each voxel precess in phase at the time of readout. When using gradient echo sequences, phase cycling is present: After an excitation pulse, the fat and water spins of a 1.5 Tesla magnet are alternatively in and out-of-phase every 2.4 ms.



In-phase and opposed-phase MR images

In-phase and opposed-phase MR images show a noticeable difference in contrast.

In in-phase images, fat and water generate an uninterrupted signal intensity in a common voxel. The transverse magnetizations of fat and water have the same orientation.

In case a gradient echo sequence generates image data in the opposed phase, a reduced signal is acquired. The sources are the transverse magnetizations that cancel one another out. This “phase nulling“ has a contrast-enhancing effect on the image. Contour artifacts occur therefore at the width of one voxel along the borders of fat and water-containing tissue.

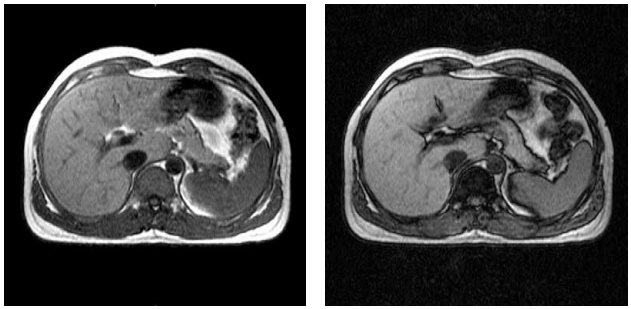
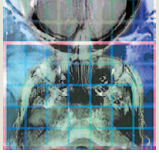


Image reconstruction in-phase (left), opposed phase (right).

What to do in case of contour artifacts?

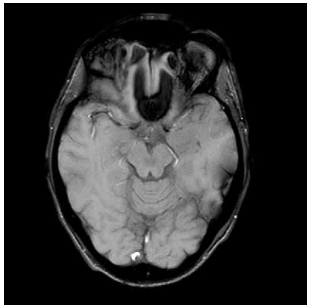
Use an echo time (TE), where fat and water spins are in phase.

0.2 Tesla: TE in-phase	36.7 ms, 73.5 ms ...
0.2 Tesla: TE opposed phase	18.4 ms, 55.1 ms ...
1.0 Tesla: TE in-phase	7.2 ms, 14.4 ms ...
1.0 Tesla: TE opposed phase	3.6 ms, 10.8 ms ...
1.5 Tesla: TE in-phase	4.8 ms, 9.6 ms ...
1.5 Tesla: TE opposed phase	2.4 ms, 7.2 ms ...
3.0 Tesla: TE in-phase	2.46 ms, 4.92 ms ...
3.0 Tesla: TE opposed phase	(1.23ms*) 3.69 ms, 6.15 ms ... * not recommended



Distortion artifacts caused by local magnetic field variations

Distortion artifacts are structures in the image that quite obviously falsify true geometric relationships.



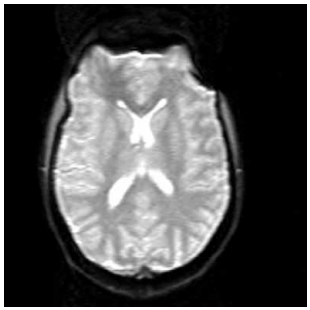
Artifacts with signal loss in the sinus region (gradient echo sequence).

Distortion artifacts in the image

The intensity of the artifact depends on local conditions. It can be shown as an increase or decrease in signal.

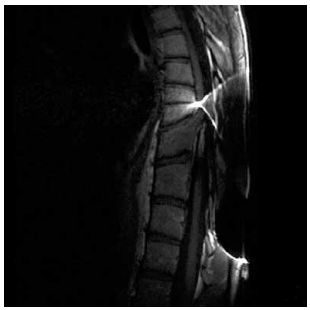
Especially susceptible are transition areas between tissue and bones or between tissue and air. Problematic areas are, for example, the paranasal sinuses, the orbits, the lungs, heart, stomach and the intestinal loops.

The distortion artifact is especially noticeable with gradient echo sequences and EPI imaging.



Distortion artifacts in EPI imaging.

Ferromagnetic objects on the patient's body or clothing (e.g., zippers) also lead to distortion artifacts.



Distortion artifacts caused through ferromagnetic objects on the patient's body.

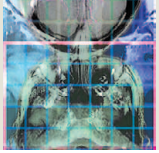
Magnetizability and field inhomogeneity

Magnetizability (SUSCEPTIBILITY) is the ability of tissue to become magnetic. At the transitions of different magnetizable tissue, local magnetic field gradients are generated and, as a result, field inhomogeneities. In most cases these are so small that they do not encourage artifacts. These inhomogeneities are completely compensated for by the spin echo technique.

With gradient echo techniques, areas of field inhomogeneity may lead to extremely heavy signal losses. The local field inhomogeneity is not compensated for.

With EPI imaging, the very low bandwidth of the sequences affect additional distortions in the phase-encoding direction.

The higher the field strength of the main magnetic field, the stronger the effect.



What to do in case of distortion artifacts?

The stronger the distortions in the MR image, the lower the bandwith of a pulse sequence or the greater the magnetic field inhomogeneity. Possible remedies are:

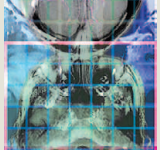
- Use of a spin echo sequence to eliminate possible signal loss by applying the rephasing 180° pulse.
- Reducing the voxel so as to reduce the differences in the magnetic field
- Shortening echo time TE to reduce the period of dephasing.
- Using sequences with larger bandwidths

Reduced to the essentials

Physically-caused artifacts are relief artifacts, as well as contour and distortion artifacts.

The sources are chemical shift and/or large jumps in susceptibility.

As compared to distortions, deformations ➔ page 140 are artifacts caused by technical errors.



Technically-caused artifacts

After motion and physics, a third influential factor, the technology used for MR, is discussed. This group of artifacts can be explained by using technical limits, such as the size of the system or the limitation of the data volume generated.

Types of artifacts

We differentiate between the following artifacts related to technology:

- Truncation artifacts
- Wrap-around artifacts
- Distortion artifacts
- Artifacts caused by RF irradiation

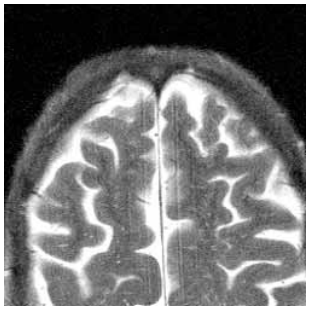
Truncation artifacts

Truncation artifacts are shown as stripes or rings in the image.

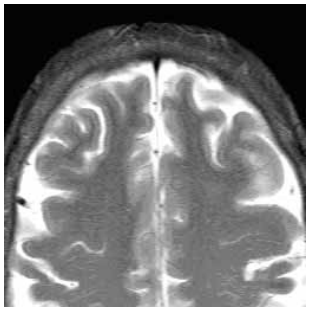
Tuncation artifacts and data sampling

Abrupt signal transitions in tissue may lead to artifacts caused by truncation. Period oscillations parallel to tissue interfaces are generated. Stripes or rings appear in the image with alternating high and low signal intensity (edge oscillations).

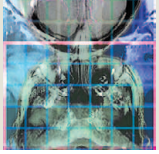
An object would be perfectly imaged, if an infinite data acquisition window were available. Through the limited time period available for the measurement, these are, however, interrupted at certain locations or not continued. Usually this does not have a negative affect on the MR image. Only highly contrast-enhanced tissue interfaces show artifacts caused by truncation.



Edge oscillation without use of a filter.



Weak filter application with minimal loss in sharpness.

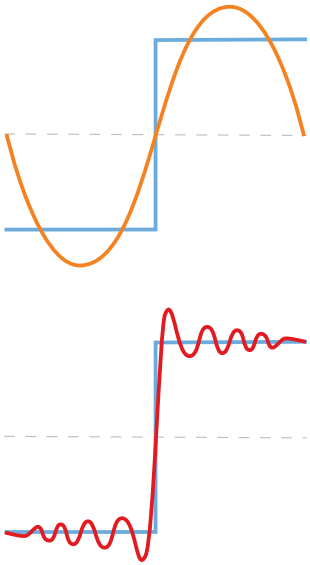


The source for edge oscillation

The edge oscillation of the artifact caused by truncation is generated while sampling the image signal. Abrupt signal transitions are simulated by curve approximations.

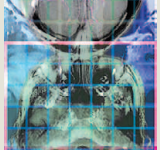
The sampling technique approximates a jump in signal intensity at the edge of the object through harmonic multiples of a sine curve. Theoretically, an infinite number of harmonics would have to be used to display a rectangle.

Through finite approximation, individual amplitude peaks occur at the edge transitions. These are known as Gibbs artifacts. They appear as oscillations of the image intensity. They are frequently known as *ringing*.



What do we do with truncation artifacts?

- Using a weak raw data filter (Hanning Filter). The strength of the filter determines the extent of loss in sharpness.
- Image reconstruction of rectangular raw data matrices automatically uses a weak filter.
- Using a larger matrix.



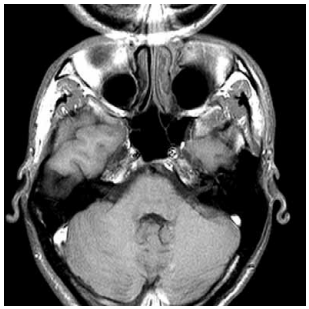
Wrap-around artifacts due to aliasing

Wrap-arounds are areas of an object that extend beyond the measurement matrix used. They are located at the opposite side as image superimposition. In most cases this artifact can be observed in the phase-encoding direction.

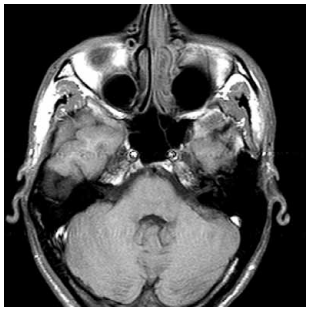
The sources for wrap-around artifacts

The size of the selected measurement field (FOV) is smaller than the object to be measured which is read out via the pulse sequence.

The tissue stimulated outside the sensitive volume of the coil contains higher or lower phase and frequency information. Through misinterpretation during Fourier Transformation (undersampling of signal), the tissue areas of the opposite side are allocated within the FOV. For this reason, the image area receives double the signal information. With a given sampling rate, only a certain maximum frequency can be interpreted correctly.



Wrap-around of the nose and back of the head located outside the image field (FOV).



Oversampling prevents wrap-around artifacts during the measurement.

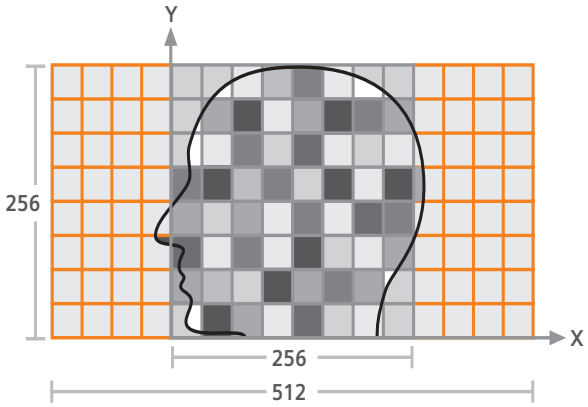
What to do in case of wrap-around artifacts?

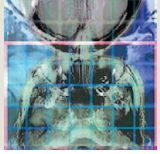
By doubling the sampling points (e.g., 512 instead of 256) during oversampling, wrap-around is successfully avoided.

The technique of oversampling is automatically used in the readout direction.

It is recommended to increase the number of sampling points in the phase-encoding direction, since the measurement time is prolonged accordingly.

Depending on the object to be measured, swapping the spatial encoding may be a suitable remedy. For transverse or sagittal slice positiong along the Y-axis, the phase-encoding direction is set by default. The X-axis is set in case of coronal slice positioning.





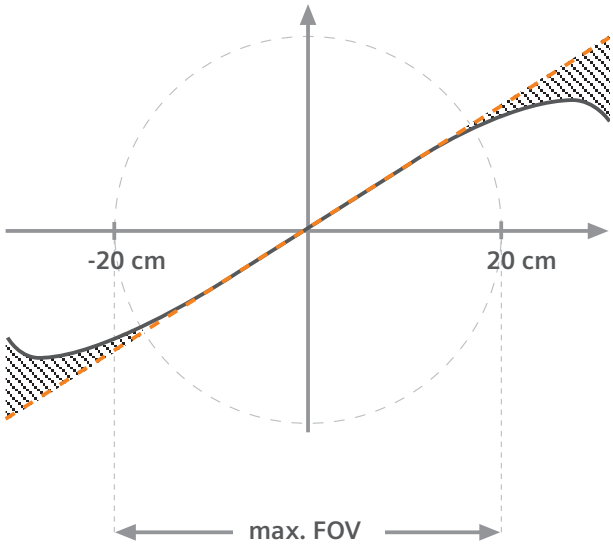
Distortions

When using measurements with an exceptionally large field of view, geometrical distortions may occur especially along the edges of MR images. These distortions are shown as deformations or distortions. The linearity limit of the gradient system is responsible for these artifacts, since it adds to incorrect image reconstruction along the image margins.

Sources for distortions

The gradient pulses encode the spatial information of an MR image. Under ideal circumstances the gradients rise linearly, however, in reality, linearity decreases at the edge of the field of view. The size of the image field that can be acquired by the gradient system, is, e.g., limited by the length of the gradient coil.

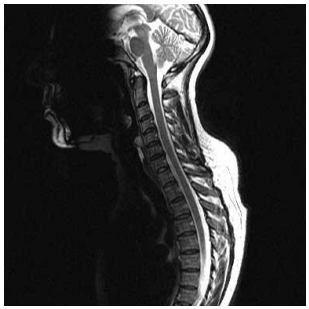
Our example shows a 5% deviation at the edge of the measurement volume measuring 40 cm in diameter. This means that all spatial information along the edge is shifted by 5% (1 cm).



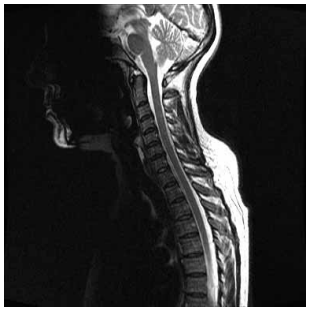
What to do in case of distortions?

By using a large FOV filter, corrections can already be performed during the measurement.

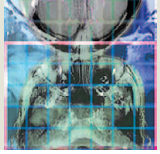
Disadvantage: The filter prolongs the reconstruction time. In addition, slice positioning in the result images is no longer distinct. For this reason, it is not acceptable.



Unfiltered original image with distortion.



Distortion correcting by using a large FOV filter.



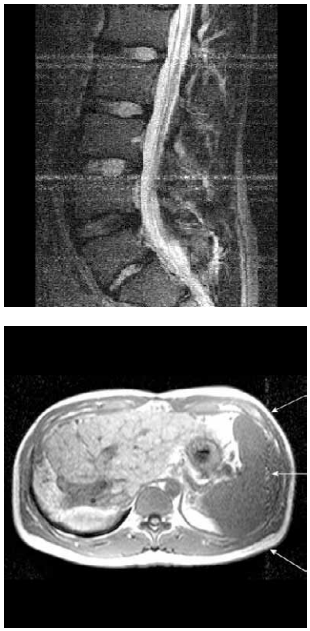
RF interferences

Image interference caused by RF irradition leads to artifacts frequently seen during the daily use of MR systems.

The sources for RF-interference

RF pulses are transmitted as well as received during the measurement. External RF fields, caused by e.g., radios, mobile telephones, electronic controls or electrical motors, emit interference signals into the MR systems that have an adverse affect on image quality.

To protect MR tomographs from the effects of radio-frequency fields, they are intalled in RF-sealed rooms (Faraday cage).
The RF-sealed room also serves to shield the environment from the effects of the RF fields generated by the tomograph.

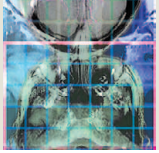


Artifacts caused by RF irradiation

What to do in case of RF irradiation?

RF interferences are usually caused by sources outside the examination room. For this reason, please ensure that the door to the room is closed properly.

After constructional changes, newly generated RF fields in the examination room may lead to interference in the RF shielding, e.g., caused by drilling holes for cables. In this case, the source for interference has to be located with care.



Summary

Artifacts may be caused by a number of different sources:
Motion, physicial conditions and technical factors.

However, many types of image interference can be prevented by carefully instructing the patient and by selecting suitable sequences and parameters. Even technical artifacts can be minimized in this way.

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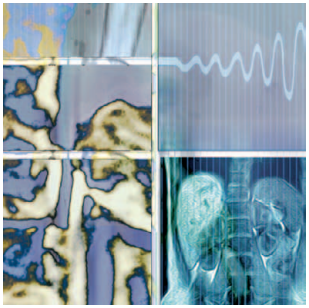

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