

The background of the slide is a complex, colorful pattern of repeating, stylized, four-lobed shapes in various colors (red, orange, yellow, green, blue, purple) arranged in a grid-like fashion. The Siemens logo is positioned in the top left corner, consisting of the word "SIEMENS" in a bold, blue, sans-serif font, with a white horizontal bar below it.

SIEMENS

Magnets, Flow, and Artifacts

Techniques and Applications of Magnetic Resonance

Magnets, Flow, and Artifacts

Magnets, Flow, and Artifacts

Techniques and Applications of Magnetic Resonance



MRI diffusion tractography
showing fiber crossings
of the right hemisphere
MPI, Leipzig, Germany

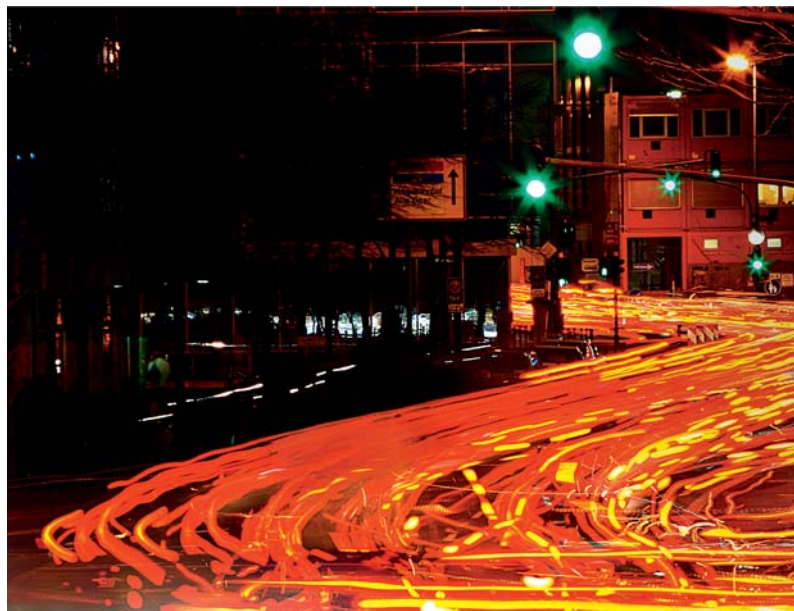


Magnets, Flow, and Artifacts

This is the second volume of *MR Basics: A Long Road Made Easy*. It covers advanced effects and techniques in Magnetic Resonance Imaging. We will be talking about spins in flow and motion, saturation, chemical shift, and functional imaging.

The volume concludes with advice for detecting and avoiding artifacts in images.

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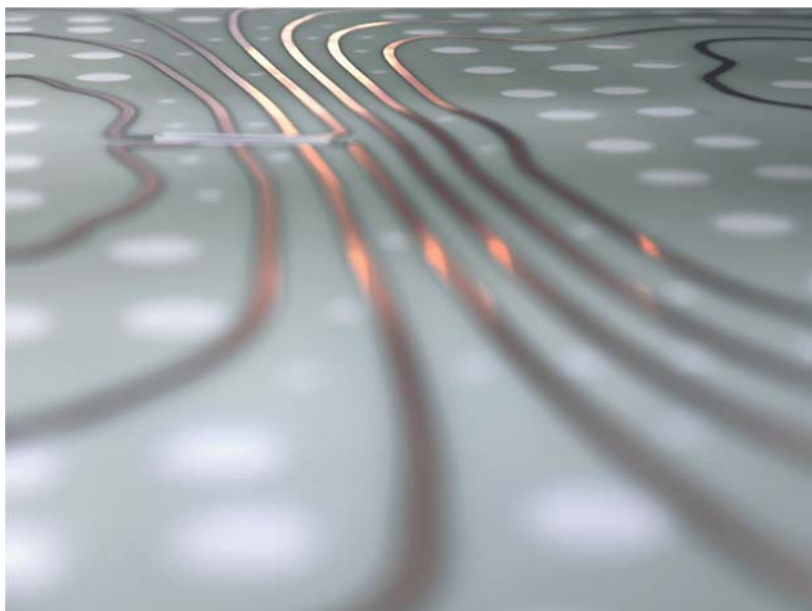


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MAGNETOM
World

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Spins in Flow and Motion

Many of the hydrogen protons in the body are not static within a volume but in motion (blood flow, CSF, diffusion, etc.). Flowing spins will either be displayed or suppressed. Relevant applications are MR Angiography and Cardiovascular Imaging.

Inflow effects

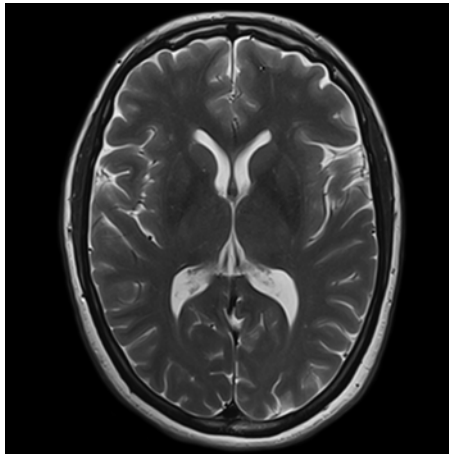
Sensitizing MR images to flowing spins

Flow in T_1 contrast

Flow refers to the flow of blood or any other body fluids. Since flow effects are the same irrespective of the body fluid in question, we will limit our example to blood flow.

Blood can be delineated as high or low in signal with respect to the surrounding tissue.

Blood has a relatively long T_1 relaxation. A standard T_1 -weighted image shows blood vessels, and other structures as well, hypointense in the image. The surrounding tissue with shorter T_1 relaxation is shown as hyperintense.

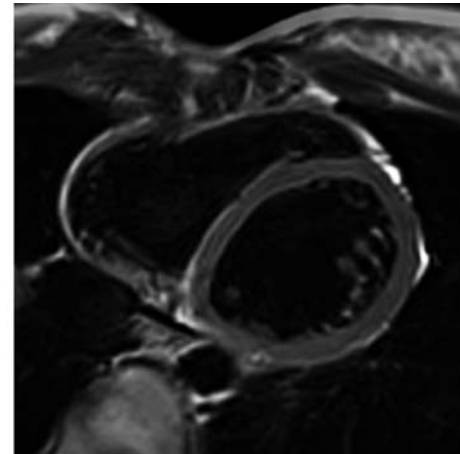
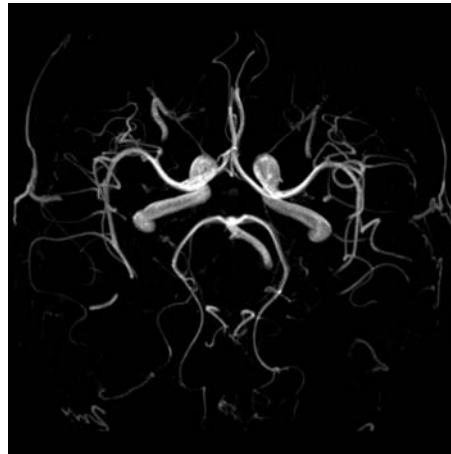


Standard T_1 -weighted image

Contrast is sensitized to the flow

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

In MR angiography, vessels are displayed using maximum intensity projection (MIP). Maximum intensity projections are computed from 3D or multiple-slice measurements and combined into MIP series.



Blood shown with high signal (on the left, head image, MIP display) and suppressed signal (on the right, heart image)

Bright blood imaging

Showing blood hyperintense in the image

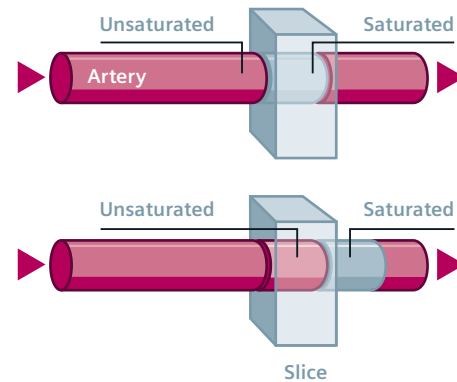
Saturating stationary spins

Unlike stationary spins, blood spins flowing into the slice are only briefly affected by the pulse sequence. The flow velocity determines how quickly they will be replaced by subsequent spins flowing in. This also affects the brightness of the blood display.

The spins are saturated by fast excitation pulses in the stimulated slice. Outside the slice, the spins are *not* excited. As soon as they flow into the slice, the excitation pulse produces 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 to the excitation pulse several times.



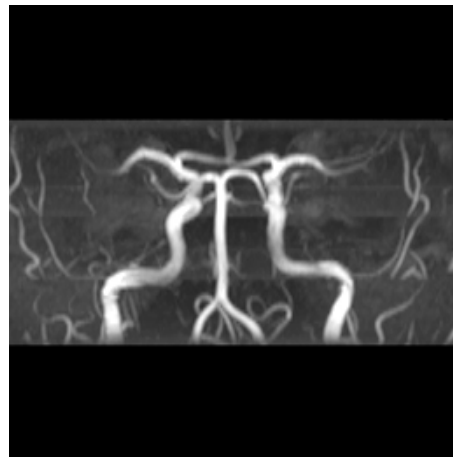
Bright blood display

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.

When does the blood signal reach this maximum level? We can calculate 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

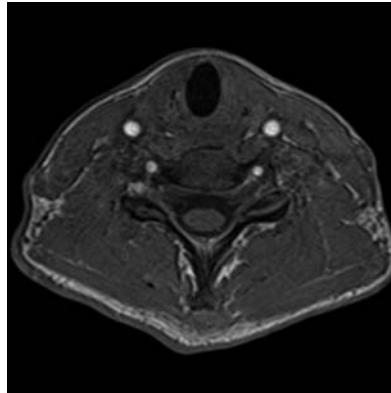
Optimizing slice orientation and suppressing undesired blood vessels

Optimal vessel course

Let us 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.

If the blood vessel lies longitudinally to the excitation slice (*in-plane*), the spins remain in the slice for a much longer time. 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.



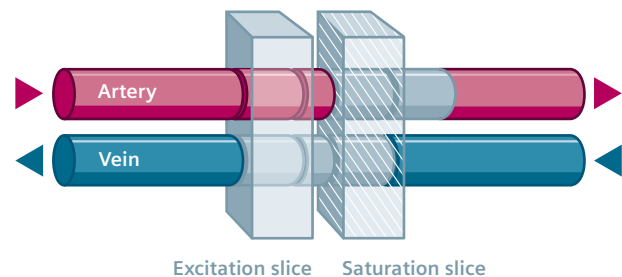
Excitation slice, orthogonal (top), used primarily for displaying carotids, and excitation slice, longitudinal to blood flow (bottom)



Artery or vein?

In most body regions, arterial and venous blood flow in opposite directions. It is an advantage that just one of the two blood flows is displayed bright.

Let us assume we would like to show the arterial flow and suppress all venous vessels. To achieve this, the vein in the excitation slice must contain saturated spins only. We position a parallel saturation slice [→ page 32](#) upstream of the excitation slice with respect to the venous inflow. Venous spins flowing through it do *not* contribute to the signal during subsequent inflow into the excitation slice. Only the artery is displayed as bright in the image.



Dark blood imaging

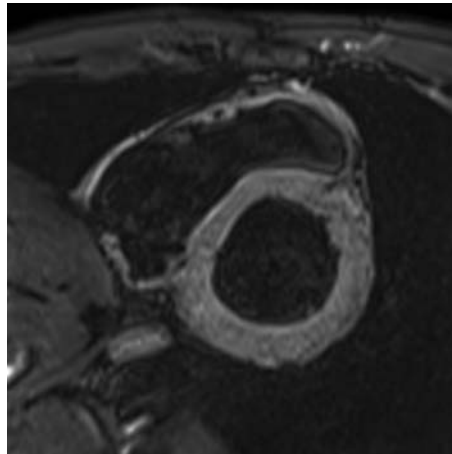
Showing blood in black

Complete nulling of the blood signal

The dark blood effect is frequently used in cardio-vascular imaging for morphological displays of the heart.

We start with preparation pulses followed by the real sequence for data acquisition. The first 180-degree 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-degree pulse reinverts the signal within the excitation slice only (for example, the slice through the heart).

The signal of the blood subsequently flowing through the slice, still inverted by the first 180-degree pulse, is acquired during the zero point of its magnetization. Only the adjacent tissue will generate a signal. The blood itself is shown in black.



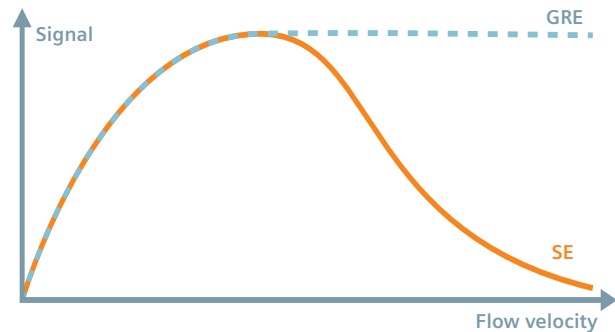
Dark blood imaging of the heart

The wash-out effect

So far, we have discussed the spins prepared before the real 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, they do not contribute to the signal.

The faster the blood flow, the fewer spins are exposed to the 180-degree pulse in the slice. The signal grows weaker. This is known as wash-out. It can create a dark-blood effect: we have a complete signal void if all of the spins excited by the 90-degree pulse have flown out as soon as the 180-degree pulse is applied.

Gradient-echo sequences do not have a wash-out effect. Additional advantages of gradient-echo sequences are their short repetition times. This makes for better suppression of the signal from stationary tissue as well as faster measurements.



Laminar flow and turbulences

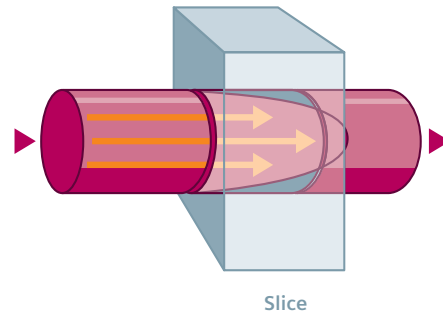
Effects caused by the phase shift of the flowing spins

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 **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, unsaturated spins in the flow direction decreases in the slice.

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

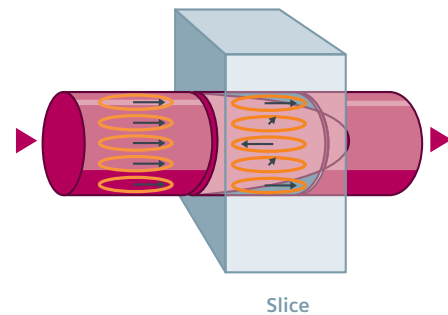


Laminar flow and spin dephasing

Let us now compare the signal of laminar and non-laminar flow in a gradient-echo sequence. During 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 18.

The faster spins move along a gradient, the stronger the phase shift.

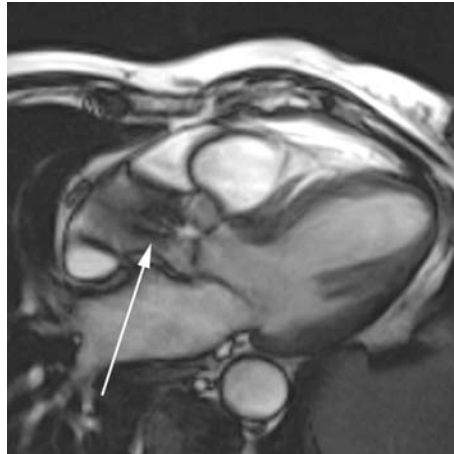
The fast flowing spins in the center of the blood vessel experience a stronger phase shift than the slower flowing spins at the edge of the vessel. The phase coherence between the spins in the blood is lost, the spins are being dephased. The signal is reduced.





Turbulences and jet effect

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

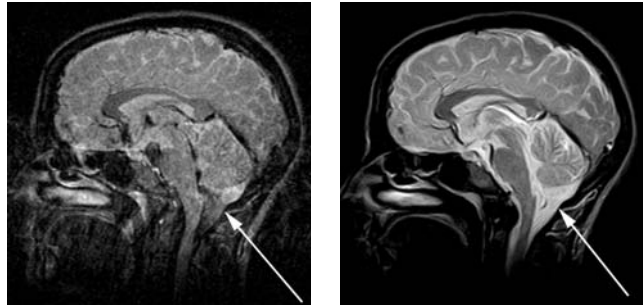
Minimizing or even eliminating interference caused by dephased spins

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 CSF on the MR image is especially noticeable. You have to compensate for signal losses in order to obtain optimal image results.

Generally, both flow images and images of body regions with a high volume of flowing spins benefit from flow compensation.

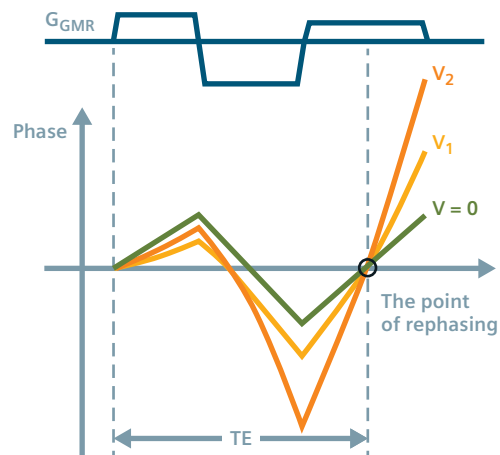


Images without (left) and with flow compensation (right). In the left image, flow leads to a region of artifacts and signal reduction in the subarachnoid space (arrow) and blood vessels. In the right image, flow compensation reduces the artifacts and enables a high signal of CSF and blood.

Flow compensation

To minimize signal loss and incorrect encoding caused by spin motion, moving and unmoving spins need to be rephased. This can be done with 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) simultaneously. Afterwards, the phase coherence is restored. The strength of the signal at the point of rephasing is the same as before dephasing.





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The inflow effect clearly delineates flowing spins from stationary spins. For this purpose, the stationary spins in the slice are saturated. Inflowing unsaturated spins produce a stronger signal. This allows us to display the course of vessels (MR Angiography).

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

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

Phase-contrast imaging

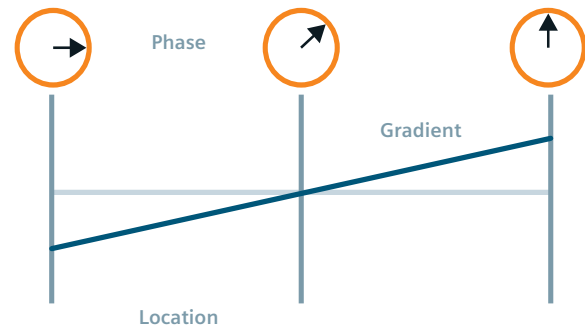
A technique that works with the phase information of spins

Moving spins and gradients

We have established that the phase of spins shift when the spins move along a gradient. When time of flight is used, this leads to interferences that have to be compensated for by additional gradients (GMR technique).

We take advantage of this effect with the phase-contrast technique. On the basis of the phase shift, we would like to determine whether and how spins are 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 spins increases linearly with a constant gradient field along the flow direction and increasing flow velocity.

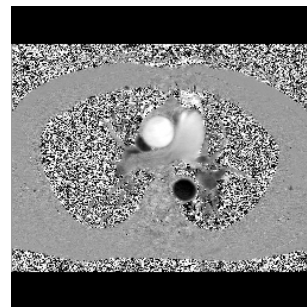
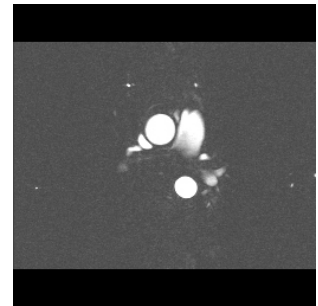
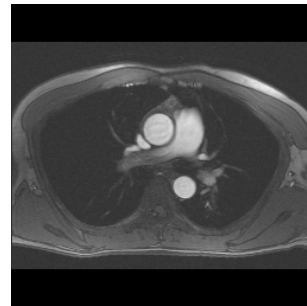


This is how phase difference is determined

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 represented in the same way as stationary spins.

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

The phase-contrast image is generated from the difference between the two sets of phase information (T_2 -weighted). There is no display of anatomical information. Instead, the pixels show the phase difference of the spins. Hyperintense pixels represent a high flow velocity in the positive direction, hypointense pixels represent a high flow velocity in the opposite (negative) direction. The mean gray value represents a flow velocity of zero, that is, stationary tissue. The scattered areas represent noise or irregular flow.



Flow-compensated image (top-left), flow-encoded image (right), and phase-contrast image (bottom)

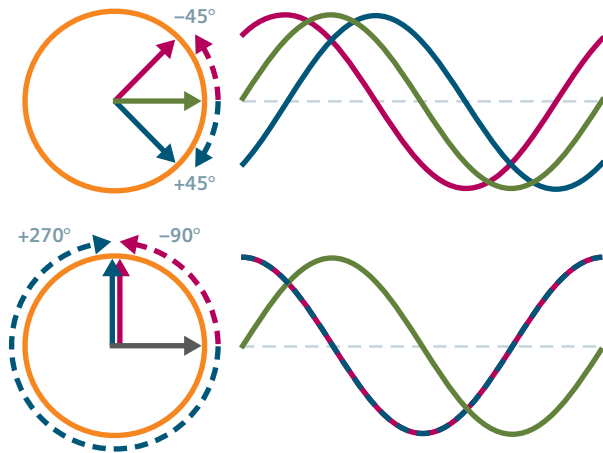
Preventing phase inversions

Velocity encoding and flow sensitivity

Phase shift and phase inversion

Phase shifts up to ± 180 degrees 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 degrees.

Let us use a phase shift of $+270$ degrees as our example. In the sinus display, this corresponds exactly to a phase shift of -90 degrees. And it is also recorded as this smaller value. A **phase inversion** occurs: The $+270$ -degree phase shift is shown as a -90 -degree phase shift.

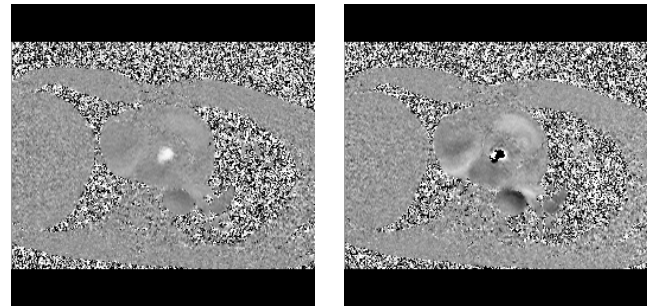


Regulating flow sensitivity

How can we prevent phase inversions? The magnitude of the occurring phase differences can be controlled via the gradient through which the spins flow. Via its parameters, for example, strength and duration, the phase difference can be increased or reduced for the same flow velocity. The gradient has a parameter-dependent **flow sensitivity**, or velocity encoding (abbreviated to 'venc').

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

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



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

Creating phase magnitude images of flow

Visualizing the phase shift

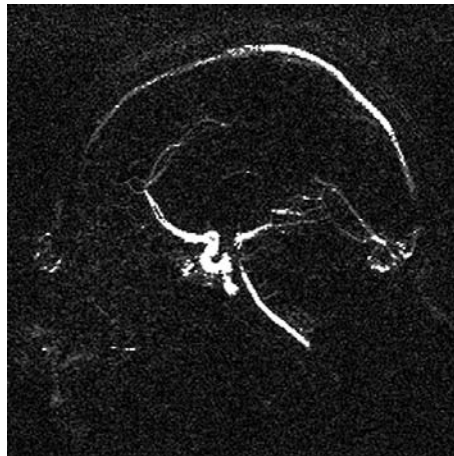
Displaying the magnitude of flow sensitivity

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.

The phase magnitude of flow is visualized in the **magnitude image**. Stationary tissue is shown in black.

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

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



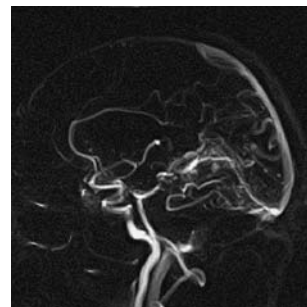
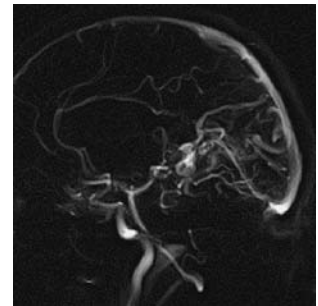
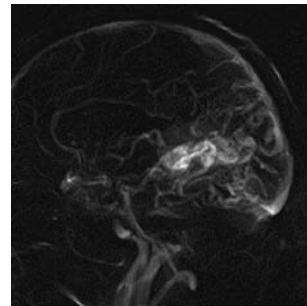
Magnitude image

Displaying spatial flow sensitivity

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

Compared with 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**.

Using a 3D sequence, the entire flow behavior in vessels can be evaluated and displayed (4D FLOW).



Magnitude sum image with
venc = 10 cm/s (top-left),
30 cm/s (right), and 60 cm/s
(bottom)

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.

Inflow effects and phase shifts offer the possibility of visualizing flowing spins either via T_1 contrast (time of flight) or phase contrast.



The background of the slide is an abstract composition. It features a dark, muted blue-grey color palette. Overlaid on this are several thin, wavy, light-colored lines that meander across the frame. Additionally, there are numerous small, semi-transparent white circles of varying sizes, some of which are arranged in a grid-like pattern in the upper left quadrant. The overall effect is a textured, organic feel.

Saturation and Chemical Shift

Frequently, the region of interest is influenced by high signal portions from tissue that is not of interest. Saturation is helpful in reducing unwanted signal.



Spatial saturation

Eliminating the signal from unwanted tissue regions

Spatial saturation techniques

Spatial saturation includes three different kinds of application:

- Regular saturation
- Parallel saturation
- Tracking saturation

With regular saturation, saturation regions can be freely positioned. With parallel and tracking saturation, saturation regions are bound to a slice group or 3D slab.

A saturation region is positioned perpendicular to the slice to be measured. Within the saturation region, a presaturation pulse is applied. When shortly after presaturation the excitation pulse of the sequence is applied, the longitudinal magnetization within the saturation region did not have enough time to relax and is still small. The signal of the saturation region is suppressed.



Example of spatial saturation



Regular saturation

Suppressing the signal of a region with a saturation pulse

Reducing motion or flow artifacts

The application range for regular saturation is demonstrated by 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 saturation, artifacts appear in the image – especially in the vertebral bodies.



With saturation, artifacts are reduced.



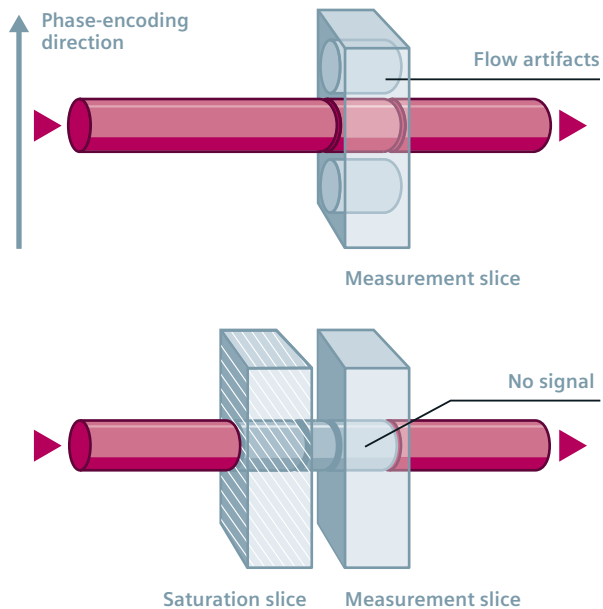
Parallel saturation

Positioning saturation regions parallel to the slice

This is how a saturation region works

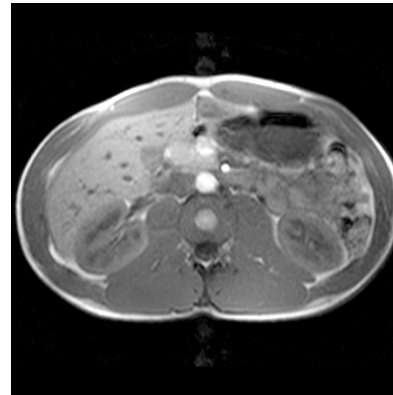
When blood vessels are imaged, flow artifacts in the form of ghosting may occur in the phase-encoding direction. Parallel saturation suppresses ghosting.

The saturation region is not in but rather outside the imaging slice. Parallel alignment with 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. Arteries and veins can be imaged separately.

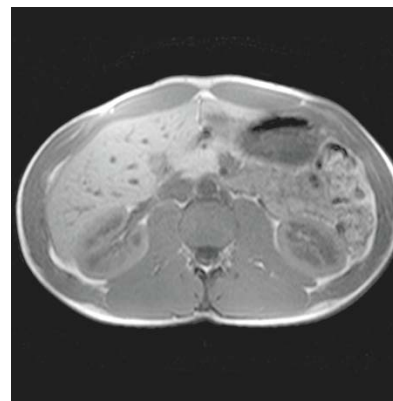


Eliminating flow artifacts

To avoid flow artifacts, a parallel saturation slice is positioned in front of and behind 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



Tracking saturation

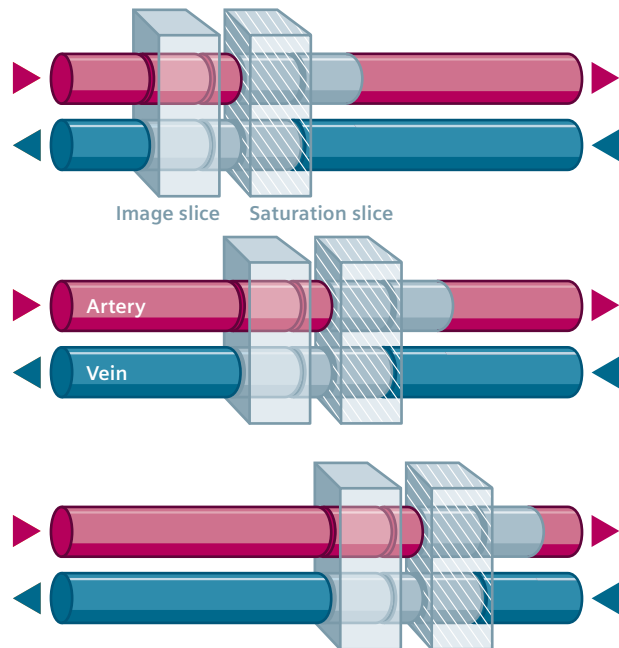
Shifting the saturation slice together with the measurement slice

Balancing the signal of flowing spins

When an image stack is acquired, 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. This may lead to ghosting in the last slices of the stack. This is prevented by tracking saturation.

In contrast to stationary parallel saturation, tracking saturation slices are positioned at one side of the image slice only. Otherwise, the entire subsequent image might be already presaturated.

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



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Spatial saturation methods are well suited for excluding motion and flow artifacts, and signal from outside the region of interest.

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



Tissue-selective saturation

Targeted signal suppression of a specific tissue or fluid

Techniques of selective saturation

We will describe three areas of tissue-selective saturation:

- Inversion recovery techniques (dark fluid) as a function of relaxation time
- Fat/water saturation, frequency-selective
- Magnetization transfer (MTC)



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



STIR

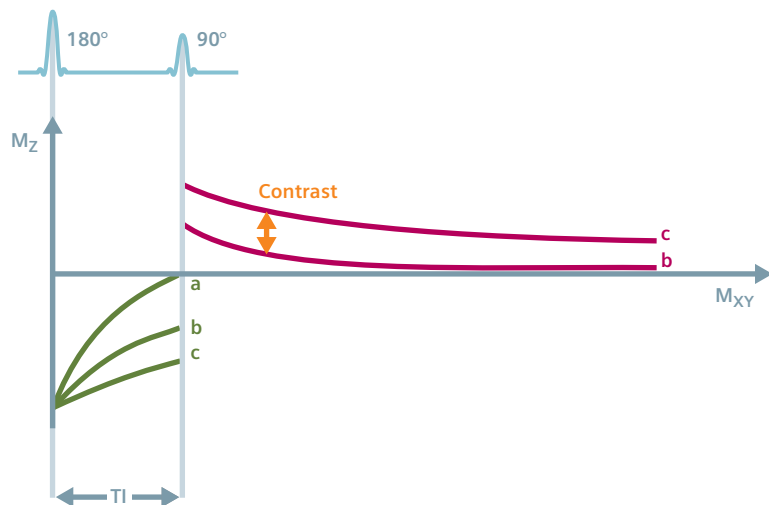
Suppressing fat signal with the inversion recovery technique

Additive T_1 and T_2 contrast

To reiterate spin-echo imaging: tissue with a longer T_1 appears darker in the image, tissue with a longer T_2 appears brighter.

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

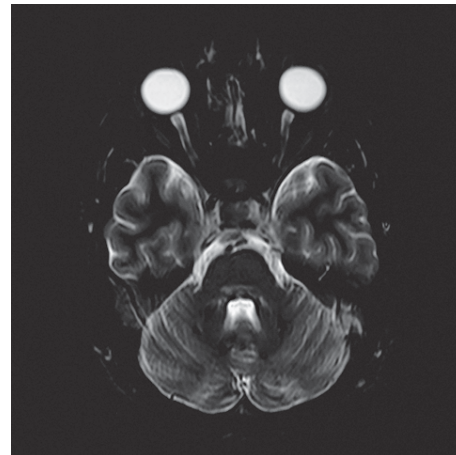
In this case, tissue with a long T_1 (b, c) shows a negative longitudinal magnetization. After the 90-degree excitation pulse, the tissue generates stronger signals (T_1 portion). With longer echo times, the contrast is further enhanced (T_2 portion). T_1 and T_2 effects work in the same direction.



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

Ideally, we would select a TI such that fat having the shortest T_1 would have just reached the zero point of the longitudinal magnetization (a). TI would have to be $0.69 T_1$. As a result, the fat signal would be suppressed (TI = 180 ms at 1.5 tesla and TI = 220 ms at 3 tesla).

TI selection depends on the main field strength.



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



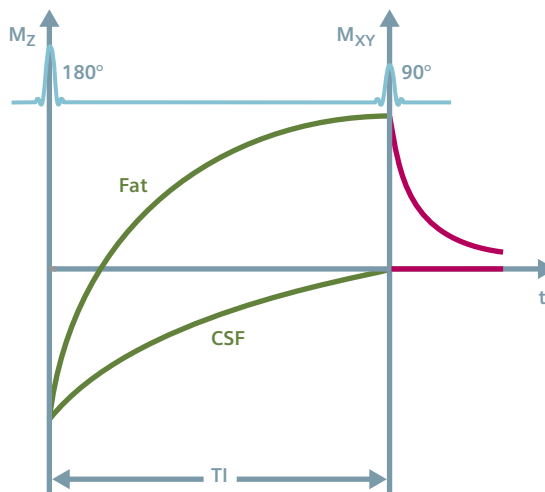
Dark-fluid technique

Suppressing fluid with the inversion recovery technique

FLAIR

Since this technique for attenuating fluid uses an inversion pulse, it is known as Fluid Attenuated Inversion Recovery (FLAIR).

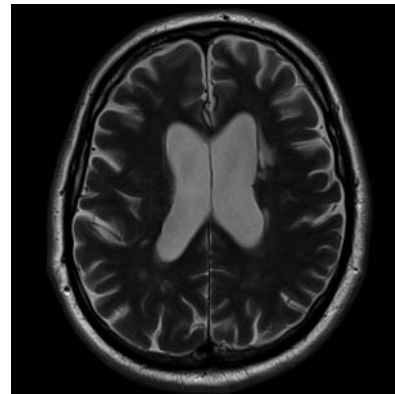
The longitudinal magnetization is inverted by a 180-degree pulse. The excitation pulse is applied when the fluid magnetization is at its zero point and does not generate a signal. As a result, CSF is displayed in black.



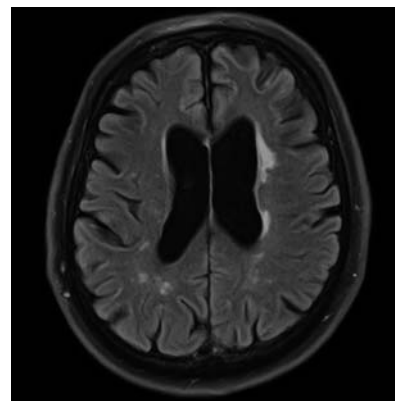
Making lesions more visible

The signal of CSF is very strong during non-saturated imaging and may superimpose lesions. With the dark-fluid technique, the signal portion of the lesion becomes more visible.

The same technique can be used with short TI inversion recovery (STIR) to suppress the fat signal.



Turbo spin-echo image



Dark fluid image



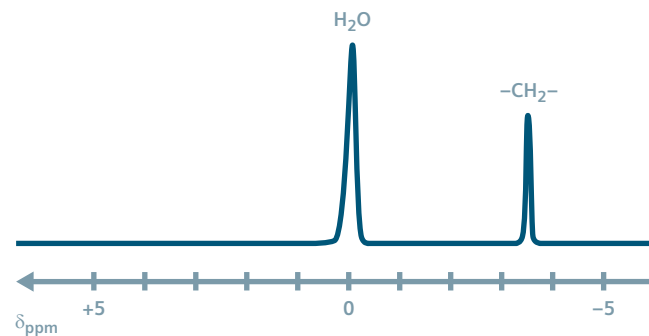
Spectral saturation

Suppressing the signal of fat or water

Chemical shift between fat and water

In almost all biomolecules, several hydrogen atoms are bound to different positions. Different positions mean different chemical and therefore usually different magnetic environments. This affects the local field strength: fat-bound hydrogen protons, for example, experience a weaker magnetic field, resulting in a lower resonance frequency.

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 ($-\text{CH}_2-$), the main component of fat. The unit used for a chemical shift is δ_{ppm} (ppm = parts per million). $\delta_{\text{ppm}} = -3.4$ means that the frequency of the methyl group is reduced by 3.4 millionths.

In spectral displays common to MR, the frequency axis is oriented from right to left.

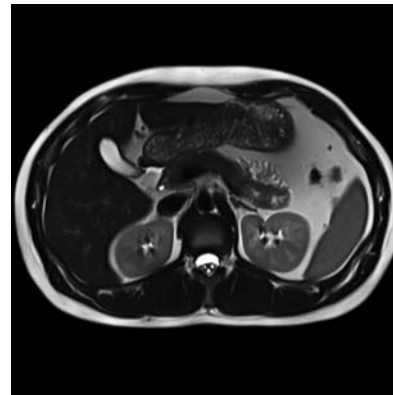
Utilizing the chemical shift

Fat is bright in many images, leading to a loss in contrast between the tissues of interest. Also, an increase in motion artifacts may be present.

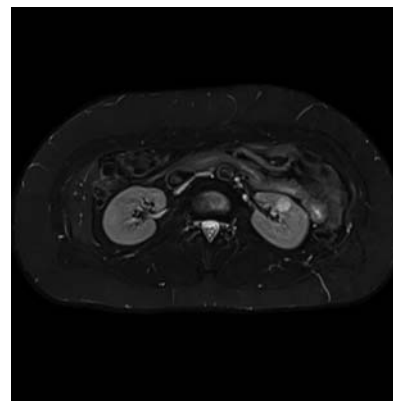
Because of the chemical shift between fat and water, a frequency-selective pulse can saturate fat. Fat spins have no time for relaxation, as a result the fat signal is suppressed.

Accordingly, the water signal can be suppressed. This is utilized, for example, in MR spectroscopy.

The **Dixon** method may be used to generate fat- and water-suppressed images simultaneously. Contrast weightings are calculated from in-phase and opposed-phase images. Pure water images are generated by adding the in phase and opposed phase, whereas pure fat images are generated by subtraction. Dixon is especially useful for eliminating field inhomogeneity effects.



Abdominal
image without
fat saturation



Abdominal
image with fat
saturation



Magnetization transfer contrast

Transferring saturation from tissue to fluids

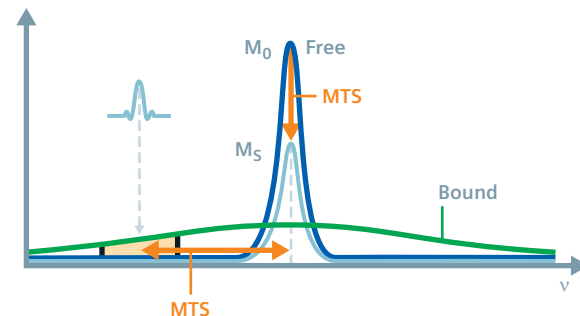
From bound to free protons

Magnetization transfer contrast (MTC) is an indirect form of saturation. The signal from certain solid tissue, for example, brain parenchyma, is reduced. The signal from liquid components, for example, blood, is maintained.

Protons bound to macromolecules 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 to 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.



Making vessels more visible

The signal is reduced in solid tissue, for example, brain parenchyma, due to the MTC technique. Blood and CSF are not affected. As a result, contrast between the two components increases. The vessels are more visible.

MTC is used in MR Angiography.



Image without MTC

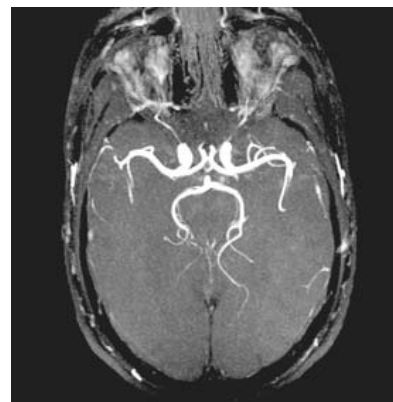



Image with MTC



Summary

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





Advanced Acquisition Techniques

MR imaging has been accelerated by optimizing the basic spin-echo and gradient-echo techniques, yielding a wealth of flexibly applicable pulse sequences. Basic reconstruction methods for parallel imaging are also discussed here.



Spin-echo variants

Advanced turbo techniques

Turbo inversion recovery

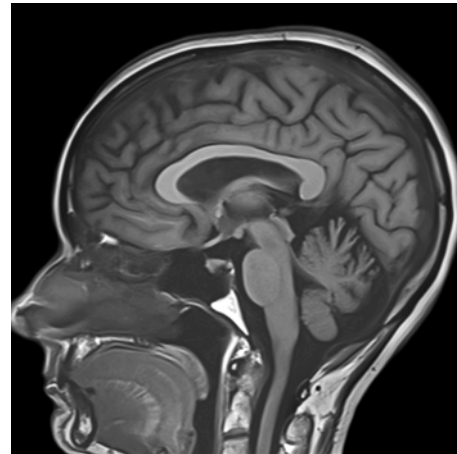
A further development of turbo spin-echo techniques (see *Magnets, Spins, and Resonances*) is turbo inversion recovery (turboIR, TIR).

TurboIR is a turbo spin-echo sequence with a long effective echo time TE_{eff} to suppress fluids. A 180-degree inversion pulse precedes the real spin-echo sequence. Contrast can be controlled by varying inversion time TI.

The turboIR sequence allows for a “true inversion recovery” display that shows the arithmetic sign of the signal (phase-sensitive IR).

Turbo inversion recovery magnitude

The Turbo inversion recovery magnitude (TIRM) technique is identical to the true IR sequence with reconstruction of the magnitude image of the signal (irrespective of arithmetic sign) and appropriate display.



Turbo IRM image

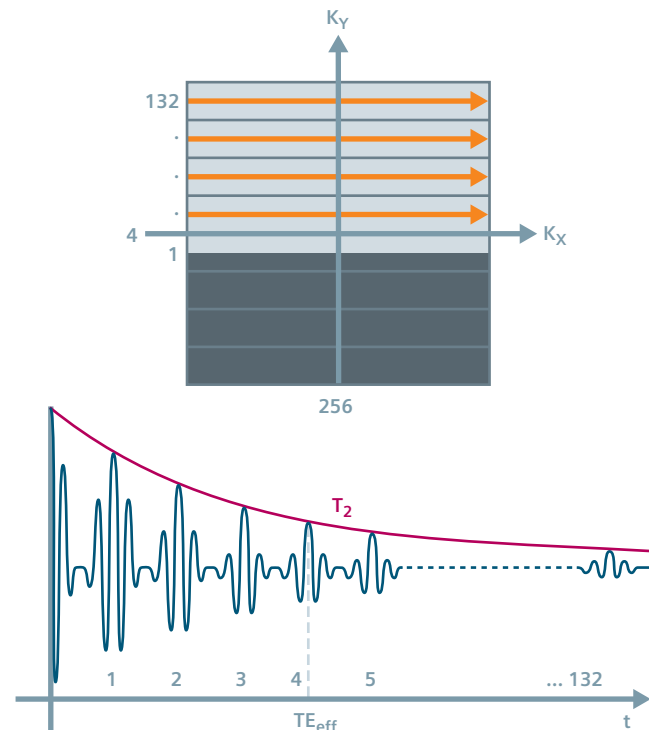
Half-Fourier technique: HASTE

Sequential acquisition of high-resolution T_2 -weighted images

One single excitation

HASTE (Half Fourier Acquisition Single Shot Turbo Spin Echo) is a turbo spin-echo technique used for sequential acquisition of T_2 -weighted images.

A single excitation generates 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 strength of the phase-encoding gradient is increased from echo to echo until the upper half of the raw-data matrix (k-space) is filled. The contrast is determined by the effective echo time TE_{eff} , that is, the echo time in the raw-data center.



Freezing movements

As only slightly more than half of the raw data is acquired, HASTE reduces the measurement time of a single slice. This keeps artifacts caused by involuntary patient movement or respiratory motion to a minimum.

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

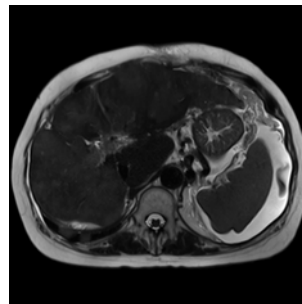
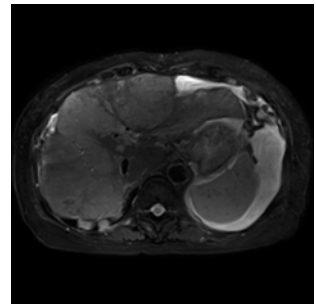
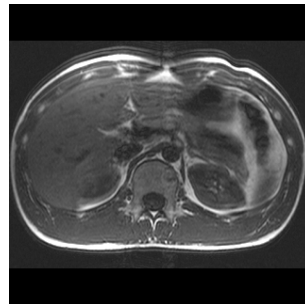


Image comparison:
Spin echo (top-left), TurboSE
(right), and HASTE (bottom)

Gradient-echo variants

TurboFLASH

Reducing motion artifacts by using short measurement times

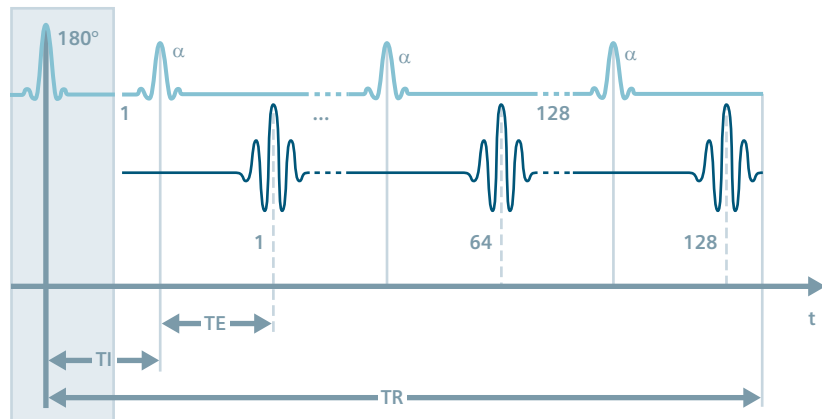
Preparation and acquisition

TurboFLASH is a FLASH sequence with echo train (see Volume 1, *Magnets, Spins, and Resonances*).

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

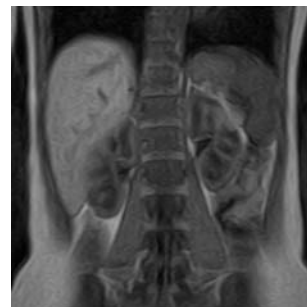
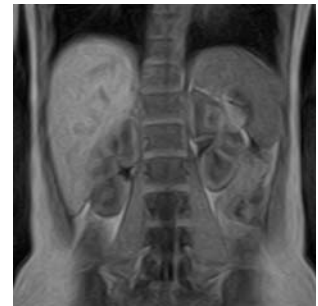
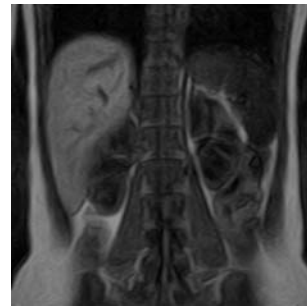
During the acquisition phase, the matrix is measured in one step with a very fast gradient-echo sequence. The distance between alpha pulses is known as the 'echo spacing.' Similar to the effective echo time of a TurboSE sequence, the inversion time TI is chosen such that the central echo fills the center raw-data row, determining optimal image contrast.

TurboFLASH allows for dynamic perfusion series after the injection of contrast agent and imaging in CINE technique.



The inversion time determines contrast

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



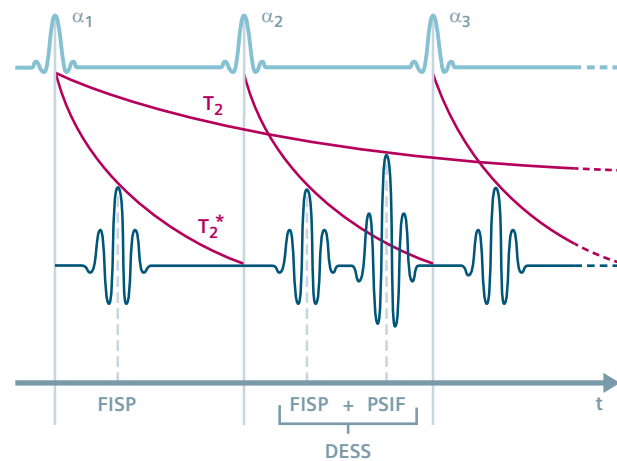
TI=50 ms (top left),
TI=400 ms (top right),
TI=800 ms (bottom left)

Dual-echo sequence: DESS

Enhancing T_2 contrast and improving SNR

Two different echoes

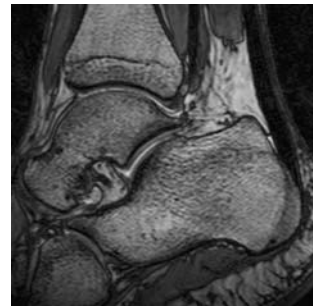
DESS (**D**ual-**E**cho **S**teady **S**tate) is a 3D gradient-echo technique where two different echoes (FISP and PSIF, see *Magnets, Spins, and Resonances*) are acquired within one repetition. During image reconstruction, the strongly T_2 -weighted PSIF image is added to the FISP image.



Enhancing T_2 contrast

DESS offers two advantages:

- The combined echoes provide for a mixed contrast: strong T_2 contrast combined with T_1/T_2^* weighting. This allows for good differentiation of synovial fluid and cartilage—an important element in orthopedics.
- Despite being not that different from a FISP sequence contrast, DESS improves SNR (because actually two images are measured).



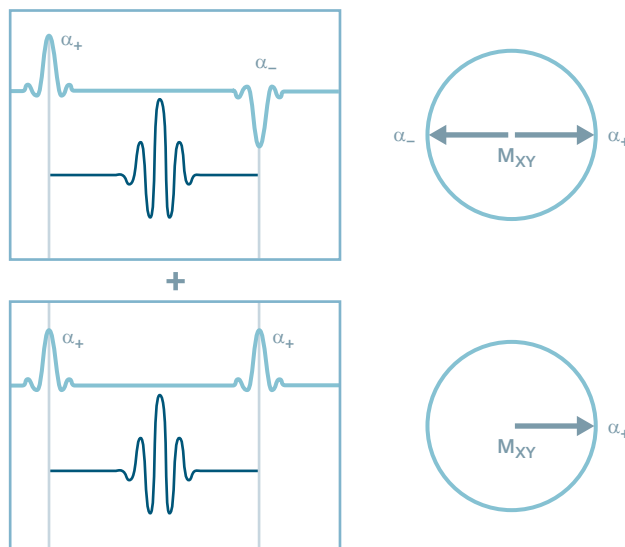
Constructive interference: CISS

Submillimeter resolution with strong T_1/T_2 contrast

Avoiding interference stripes

CISS (Constructive Interference in **S**teady **S**tate) is a strong T_2 -weighted 3D gradient-echo technique with high resolution, where two acquisitions with different excitation levels are performed internally and subsequently combined.

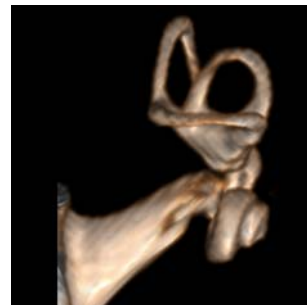
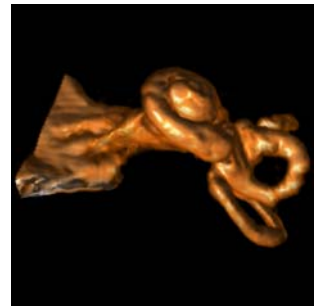
CISS combines two TrueFISP sequences. TrueFISP provides the highest signal of all steady-state sequences. During one measurement the phase angles of the RF pulse are alternated (α_+ , α_-), in the other measurement they are *not* alternated (α_+ , α_+). The sum images are T_1/T_2 weighted and may include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



The advantage of 3D CISS

3D CISS 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 example demonstrates one of the advantages of the 3D technique: With thin slices, the higher SNR allows for excellent detail detection of anatomical structures.



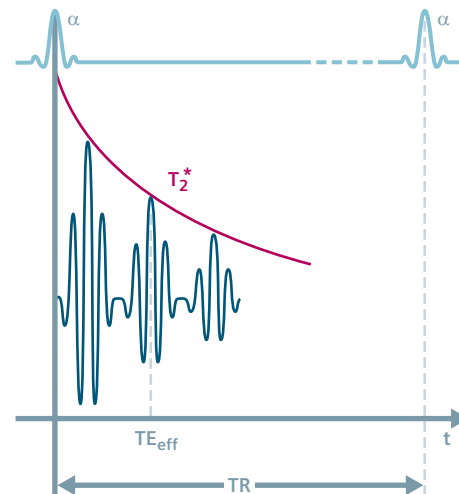
Multiecho sequence: MEDIC

Reducing flow artifacts and effects caused by chemical shift

Combining several images

MEDIC (**M**ulti-**E**cho **D**ata **I**mage **C**ombination) acquires multiple echoes in a single measurement which are then combined into an image. This results in a higher SNR per time period. Flow effects are compensated with each echo.

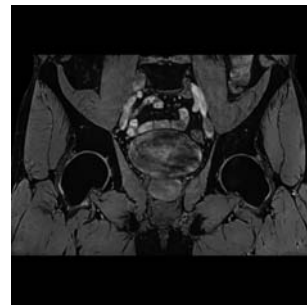
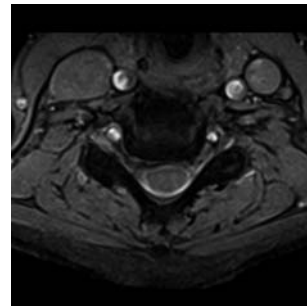
Since images with different echo times are combined, the new images show a mixed T_2^* contrast. For this reason, only an effective echo time TE_{eff} can be provided as echo time.



Reducing chemical shift artifacts

MEDIC not only minimizes flow artifacts. The sequence also reduces artifacts connected with the chemical shift.

Typical applications include the cervical spine and orthopedic requirements.



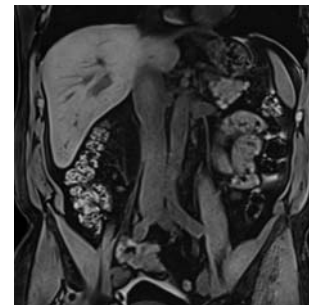
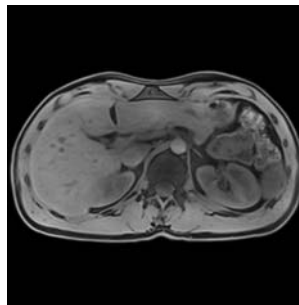
Breathhold sequence: VIBE

T₁-weighted high spatial resolution

High image quality despite short measurement time

VIBE (Volume Interpolated Breathhold Examination) combines two techniques: the half-Fourier technique accelerates the measurement of partitions in the slice-select direction. The 3D FLASH measurement is accelerated by interpolating the measurement points.

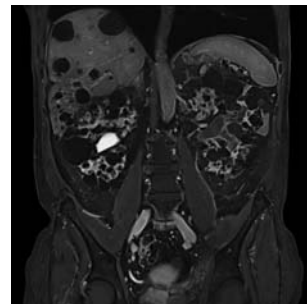
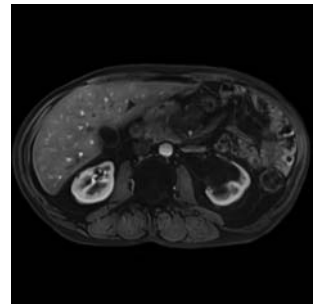
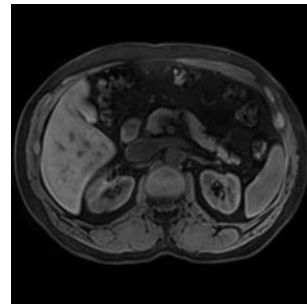
Although GMR flow compensation is omitted, an angiographic image effect is obtained based on extremely short echo times. VIBE always uses fat saturation. In dynamic contrast agent studies, the fast VIBE sequence provides for timely precision in the acquisition of vessels in the arterial and venous phase,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



Improving partial volume effects

The VIBE sequence reduces partial volume effects.

Areas of application: breath-hold technique of the abdominal region and the thoracic and pelvic regions.



Turbo gradient spin echo (TurboGSE)

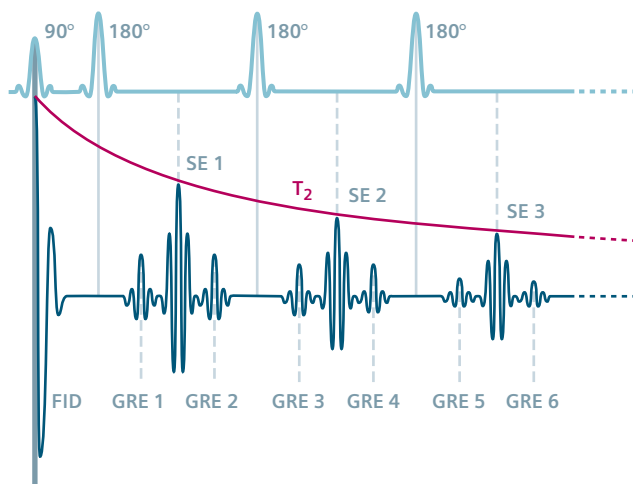
Pure T_2 contrast for high-resolution imaging or extremely short measurement times

Using additional gradient echoes

TurboGSE (turbo gradient spin echo) is an expansion of the turbo spin-echo technique. A TurboGSE sequence generates additional gradient echoes before and after each spin echo, using dephasing and rephasing gradient pulses switched accordingly.

How do the spin echoes and gradient echoes fill into the raw-data matrix (k-space)? The spin echoes provide the center segments and ensure contrast. The gradient echoes determine the resolution of the outer segments.

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, i.e. fat is darker compared to TurboSE sequences.



Faster than TurboSE

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, i.e. fat is darker compared with TurboSE sequences.

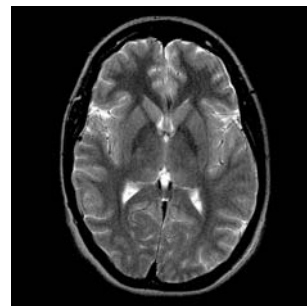
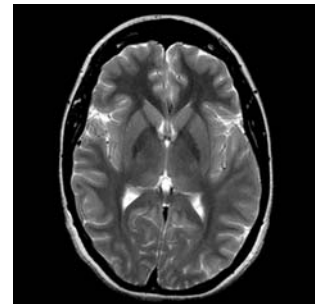
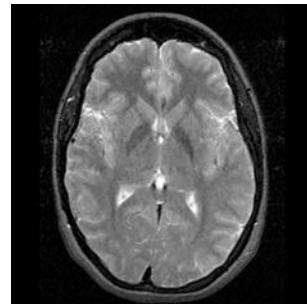
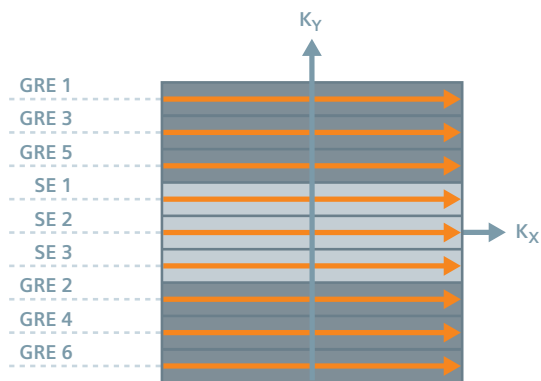


Image comparison:
T₂ spin echo (top-left, measurement time of 7 minutes),
T₂ TurboSE (right, 8 seconds),
and T₂ TurboGSE (bottom, 6 seconds).

Segmented EPI

Reducing off-resonance effects and distortion artifacts

Single-shot EPI sequences (see Volume 1, *Magnets, Spins, and Resonances*) are very sensitive to off-resonance effects. Off-resonance means that spins outside the excited slice contribute to the MR signal and may lead to image artifacts.

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

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

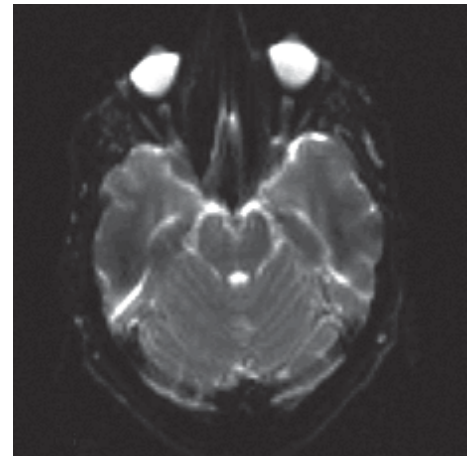
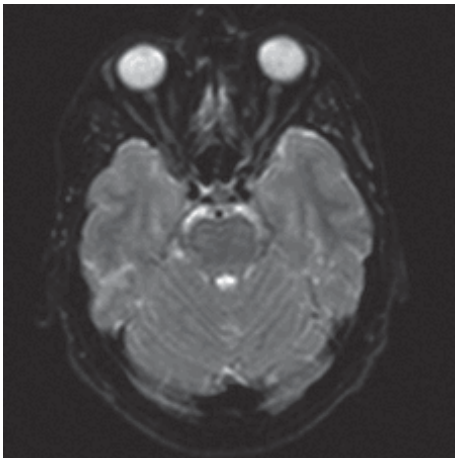


Image comparison: Single-shot EPI image with distortion artifact.



The segmented EPI image shows considerably less distortion in the area of the eyes.

Methods for parallel acquisition

Imaged-based parallel acquisition (SENSE)

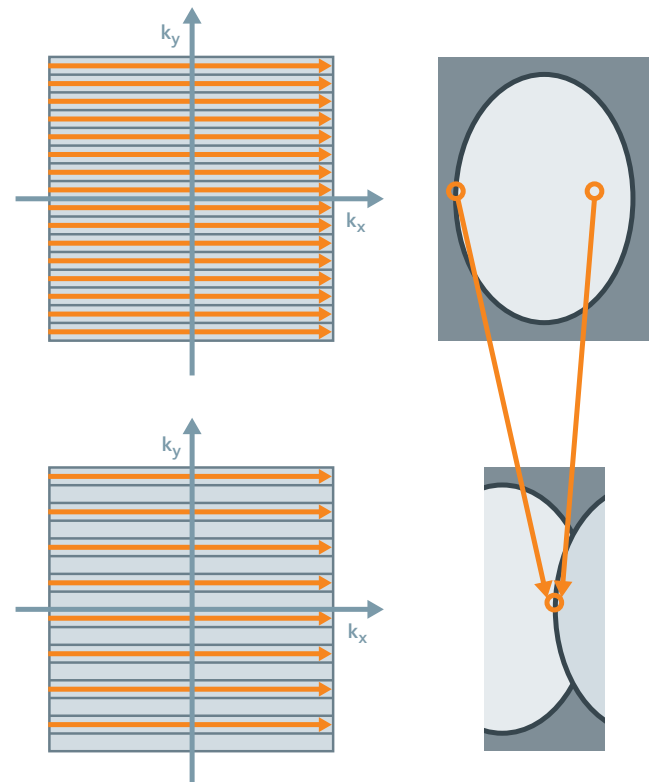
Sensitivity encoding for fast imaging

Reduce and overfold

The SENSE algorithm (*Sensitivity Encoding*) reconstructs the MR image from the image data of the individual coil elements after the Fourier transform. For principles of parallel imaging, see *Magnets, Spins, and Resonances*.

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

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



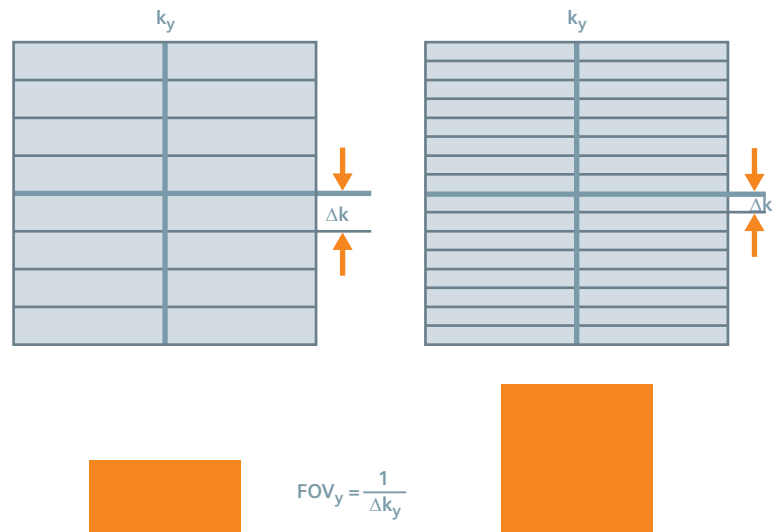
The reduced image of a phantom shows periodic overfolding in the areas outside the field of view.

Field of view, resolution, and sampling rate

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

The sampling rate is the inverse of the field of view:

$$\Delta k = \frac{1}{FOV}$$

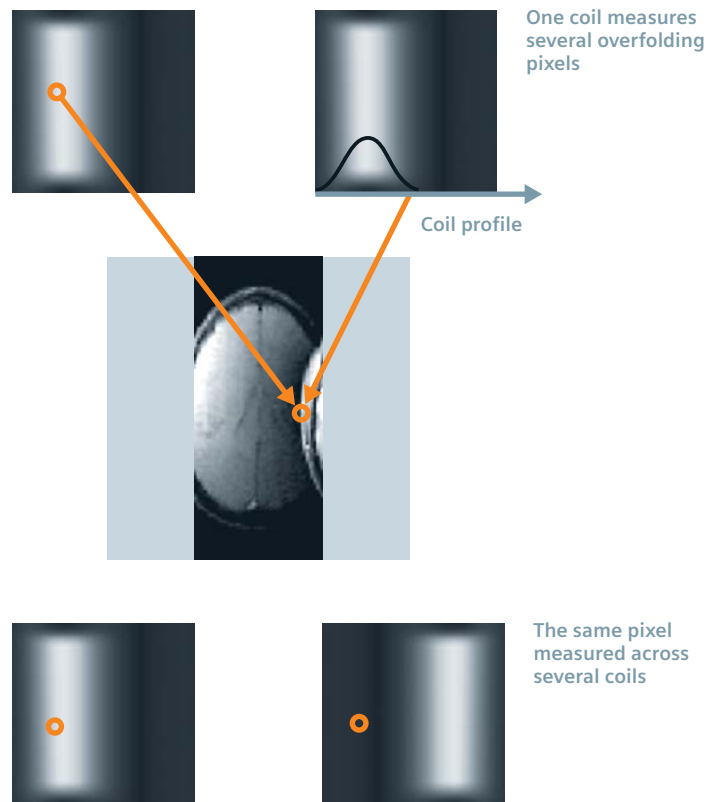


Overfolding and unfolding

SENSE differs from a folded up transparency in that a coil element is *not homogeneous*, but rather has a spatial sensitivity profile, the overfolding pixels are not the same in the images of the individual coils. They are weighted with the spatial coil sensitivity.

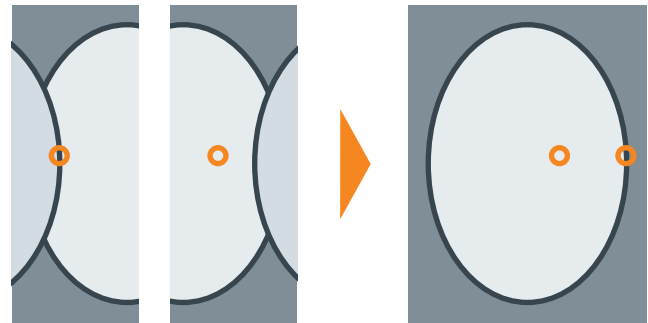
The question remains as to how we obtain the unfolded total image from an overfolded image.

If we have only one image with overfolding, we cannot unfold it properly. Where possible, we can avoid overfolding by **oversampling**. But if we acquire several overfolding images in parallel across several coils, we can reverse overfolding by using a dedicated image reconstruction algorithm.



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

For more information about overfolding and oversampling refer to section *Wrap-around artifacts (aliasing)* of Chapter 5, *Detecting and Avoiding Artifacts*.



Unfolding

Raw-data based parallel acquisition: GRAPPA

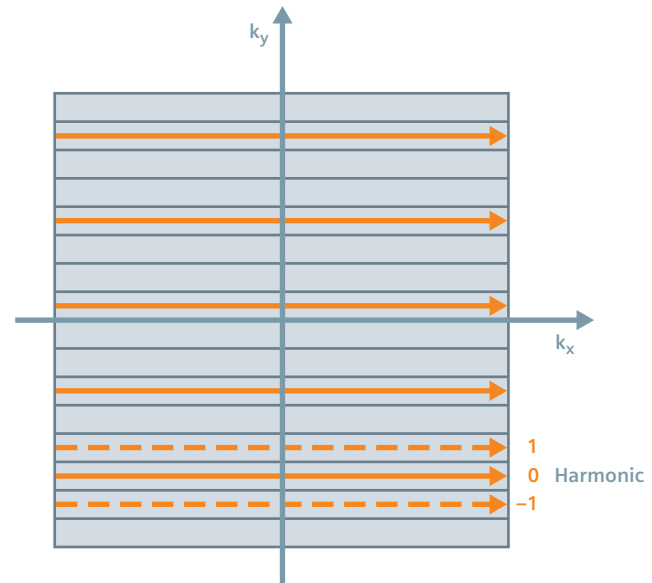
Parallel reconstruction prior to the Fourier transform

Harmonics in k-space

Unlike SENSE, GRAPPA (*Generalized Autocalibrating Partially Parallel Acquisition*) reconstructs the MR image from raw data. A number of phase-encoding steps are skipped. The missing raw-data rows are completed using a certain trick.

The values in the k-space are spatial frequencies that correspond to stripe patterns that make up the image. These stripe patterns are structures recurring periodically in the object to be measured. The phase encoding generates these patterns from the spin phases.

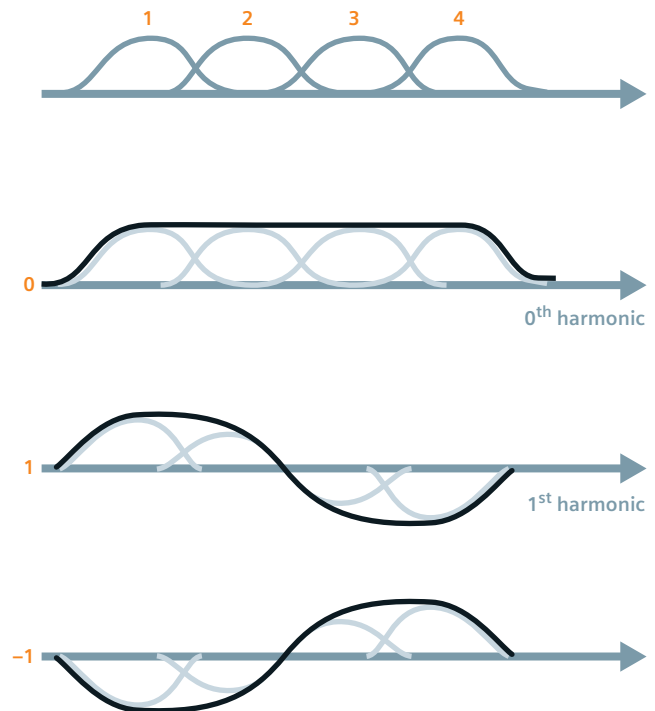
Let us assume a receiver coil has a coil profile corresponding exactly to the wave pattern. The respective phase-encoding step would then be unnecessary. Instead of phase encoding one could theoretically, if technically possible, increase a wavy coil profile step by step. This would have the same effect.



The coil profile trick

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

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






CAIPIRINHA

CAIPIRINHA is a parallel imaging technique, which modifies the appearance of aliasing artifacts during data acquisition to improve subsequent parallel image reconstruction.

CAIPIRINHA significantly reduces the measurement time of breath-hold measurements without affecting image resolution, coverage, or contrast.





Functional and Quantitative Imaging

In functional MRI, the information displayed not only relates to anatomical structures but also to functional and quantitative measures. These can be overlaid as graphs and highlights in the anatomical image, enhancing diagnostic value.

Diffusion-weighted imaging

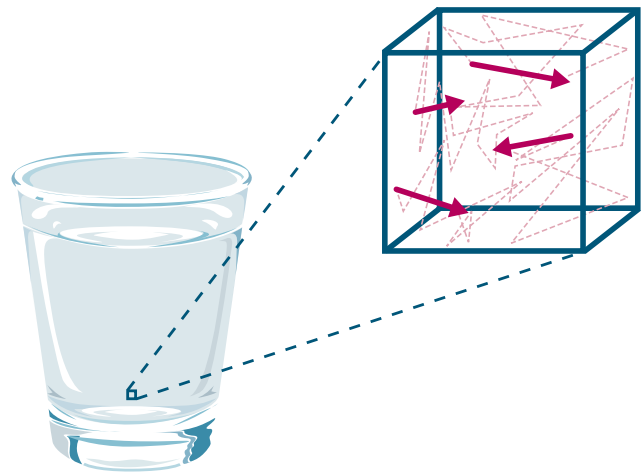
Monitoring the molecular motion in tissue

What is diffusion?

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

You can watch this phenomenon by dipping a teabag into a glass of cold 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).



Diffusion toward lower concentration

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

Differences in concentration exist within all tissues, for example, areas that are high in nutrients versus those that are low in nutrients. These differences enable molecules to diffuse in a specific direction: the direction of lower concentration. If you monitor molecules for a longer period of time, you will see a linear net shift (colored arrows).

Apparent diffusion

Making the mobility of molecules visible

The diffusion coefficient

The average net shift of the molecules depends on the type of tissue. This is described by 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

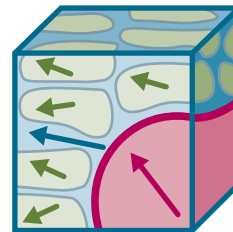
CSF	approx. $3 \times 10^{-3} \text{ mm}^2/\text{s}$
Gray matter	approx. $0.8 \times 10^{-3} \text{ mm}^2/\text{s}$
White matter	as a function of direction $0 - 1.1 \times 10^{-3} \text{ mm}^2/\text{s}$

As a function of direction → page 88

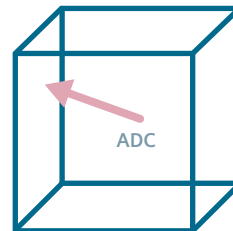
Apparent diffusion coefficient (ADC)

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 expressed as **apparent diffusion coefficient**, or ADC. It is the *time-averaged* diffusion coefficient of the voxel.



Volume element with different tissue compartments showing different diffusion coefficients

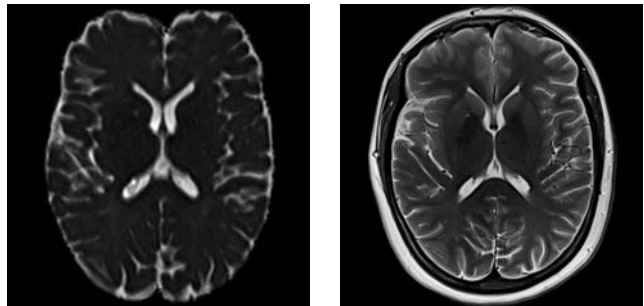


Volume element with average diffusion coefficient in a certain direction

The diffusion image and 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 hypointense pixel represents a voxel with a low ADC and consequently low diffusion. A hyperintense pixel represents a high ADC and therefore high diffusion.



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



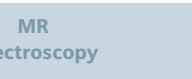
Diffusion-weighted
imaging



Perfusion
imaging



BOLD
imaging



MR
Spectroscopy

The principle of diffusion weighting

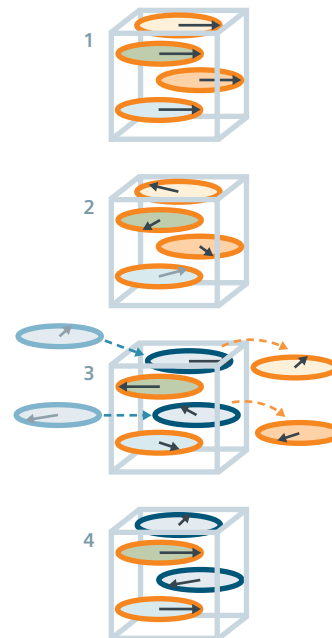
This is how MR diffusion images are generated

Attenuating the signal of diffusing spins

Diffusion is determined with two opposing gradient pulses, switched briefly one after the other. This is the 'bipolar diffusion gradient.' We start with a coherently precessing spin ensemble within a voxel (1). It is completely dephased by the first gradient pulse (2). Subsequently, several spins diffuse out of the voxel and are replaced by spins from adjacent voxels (3).

The second gradient pulse completely dephases stationary spins. Diffused spins of an originally different spin phase, however, cannot be fully rephased by a negative gradient pulse. The signal of the new spin ensemble is reduced (4). This signal attenuation creates a new contrast: **diffusion weighting**.

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



The b-value controls diffusion weighting

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:

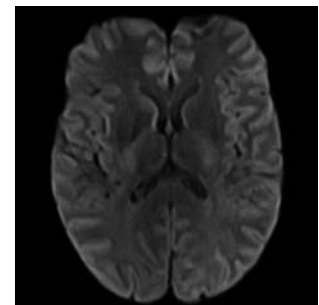
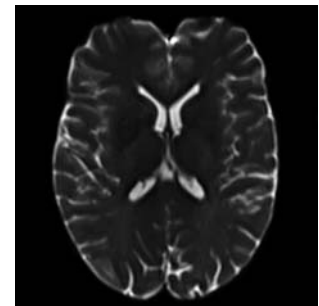
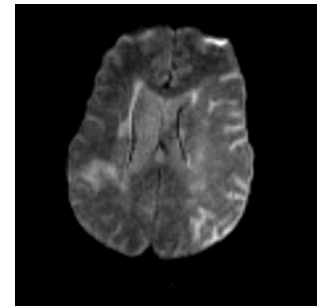
- strength (amplitude)
- duration
- time interval between the two single pulses

The increase in signal decay is expressed by the strength of diffusion weighting (**b-value**, in units of s/mm^2). A higher b-value leads to a more pronounced signal decay with increasing diffusion.

The value $b=0$ indicates that diffusion weighting will not be performed (T_2 comparative image).

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 $150 \text{ s}/\text{mm}^2$. In this range, perfusion spins show a complete signal loss due to dephasing. In addition to the reference image ($b=0$), DW images with b-values of $500 \text{ mm}^2/\text{s}^2$ and $1000 \text{ mm}^2/\text{s}^2$ are generated.

Diffusion-weighted images:
 $b=0 \text{ s}/\text{mm}^2$ (top),
 $b=1000 \text{ s}/\text{mm}^2$ (bottom).



Comparing diffusion-weighted images with ADC maps

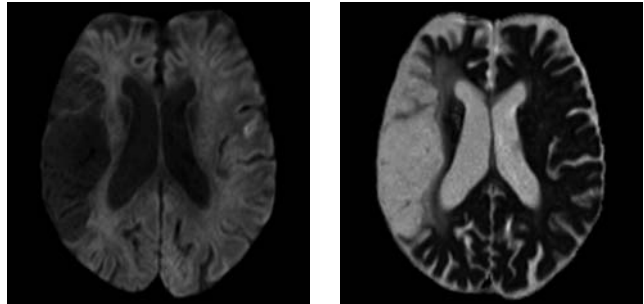
Diffusing disorder differently visualized

Different contrast, same disorder

ADC maps show the diffusion coefficient as functional information and do not contain anatomical signal portions. Diffusion-weighted (DW) images additionally show anatomical information since they contain T_2 -portions. Yet both images show the same diffusing disorder.

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

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



Diffusion-weighted image with $b = 1000$ (left) shows an old diffusion disorder in the left hemisphere. In the ADC map (right) this area is brighter.

Anatomical issues

Why aren't we satisfied with anatomical DW images? Diffusion-weighted images contain signal portions that come directly from the tissue irrespective of the 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 eliminate the possibility of an erroneous diagnosis on the basis of a T_2 shine-through effect. However, there are additional complications, notably the anisotropy of diffusion → [page 88](#).

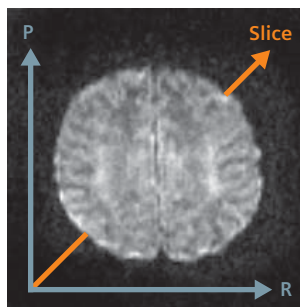
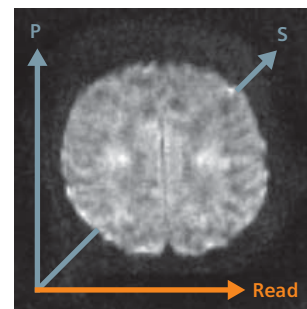
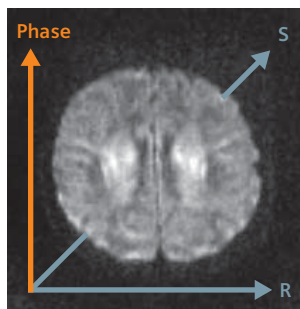
Directional diffusion

Different signal intensities depending on the orientation of the diffusion gradient

Utilizing diffusion anisotropy

The orientation of nerve fibers provides an example of directional diffusion. A myelin sheath surrounds the 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.

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



Alignment of the diffusion gradient in the phase-encoding direction (phase, left), read-out direction (read, right), and slice-selection direction (slice, bottom). All three images show the same slice, but different contrast due to the orientation of nerve fibers.

Diagnostic value of averaged diffusion

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

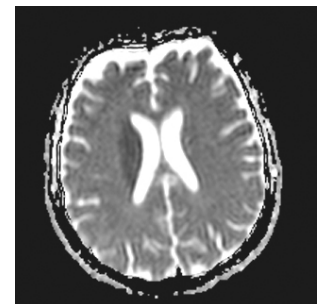
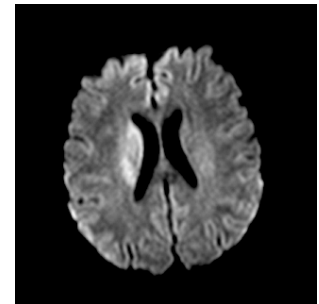
The **trace image** is the average of the three measurements in the orthogonal directions. This is the easiest way of generating diffusion images that are independent of anisotropy.

An averaged ADC map is generated in 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 with the diffusion axes.

ADC map or ADC trace? What is the best diffusion display? Due to its higher diagnostic value, the averaged ADC map has almost completely superseded the standard ADC map.

Trace image (top) and
averaged ADC map (bottom)





Diffusion tensor imaging

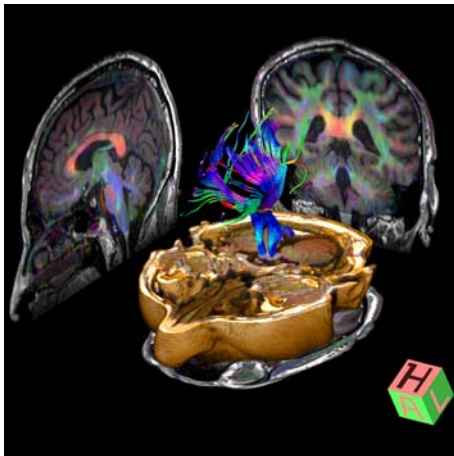
The diffusion tensor

An anisotropic magnitude can be mathematically expressed as a tensor. This is, so to speak, the next dimensional leap after scalars 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 strength in the three orthogonal directions. The sum of these three diagonal elements (*trace* of the tensor) results in the 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 circumstances, you only need diffusion gradients in six directions to fill the entire diffusion tensor. In order to display complicated anisotropies, you will need to apply as many as 12 or more different diffusion gradients (multi-directional diffusion weighting, MDDW).

The directional dependencies of the diffusion tensor image can be displayed in direction-dependent colors or as a kind of averaged display, known as the fractional anisotropy map (**FA map**).



DTI images superimposed with anatomical images and reconstructed fiber tracking in 3D

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

Diffusion tensor imaging (DTI) is all about direction. This technique helps delimit or selectively display the fiber connections of white matter or the individual core areas of deep gray matter.

Perfusion imaging

Visualizing changes or delays in microvascular blood flow

First pass of a bolus

Perfusion is the vascular transport of nutrients for supplying the cells in the capillary bed of tissue. Perfusion images are created, for example, by single-shot EPI sequences. This enables rapid image acquisition with high sensitivity. For example, we use perfusion in the brain. Perfusion also takes place in the heart or any other tissue.

Perfusion is traced by injecting a contrast agent (CA) intravenously. 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. If perfusion is satisfactory, the contrast agent is finely distributed in the tissue via the capillaries and will be washed out by the blood that follows.



Susceptibility contrast

During first pass, the signal does *not* increase as might be expected. Instead, the signal *decreases* significantly due to the T_2^* effect of the contrast agent.

The reason for the reduced signal is the change in T_2^* . This change is the result of the difference in magnetization between the intravascular space filled with contrast agent and the surrounding tissue, the difference in susceptibility of both areas. Dynamic shows in which region contrast agent does *not* accumulate. This imaging method is also known as dynamic susceptibility contrast (DSC) imaging.

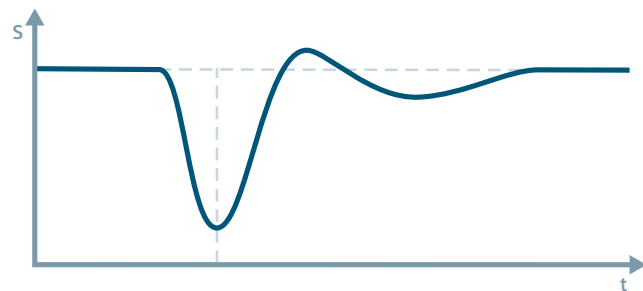


Global bolus plot

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.

The global bolus plot provides only a general description 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.



Global Bolus Plot (GBP)

In biological terms, perfusion is the flow of nutrients through the capillaries. The flow begins with the vascular transport (from the arterioles) into the capillary bed. This is followed by the diffusion of the molecules of the nutrients through the cell membranes into the cell to be supplied. Reperfusion transports the waste product from the cells to the venous and lymphatic 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 perfusion.

Besides the Dynamic susceptibility contrast (DSC) technique for perfusion evaluation, there are non-contrast methods available. Arterial spin labeling, for example, measures and evaluates the inflow of RF-excited blood.

Perfusion maps

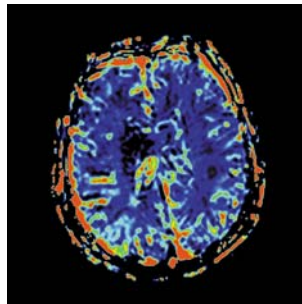
Individual cards and maps generated for every slice measured

Blood volume and blood flow

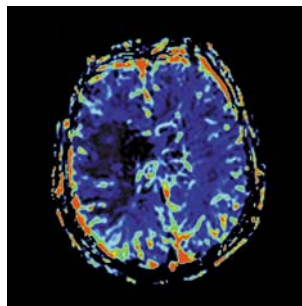
The regional blood volume and blood flow in the brain are the two main parameters for perfusion imaging.

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

Mean transit time

The **mean transit time** (MTT) is the ratio of CBV to CBF, a sensitive index of decline in perfusion pressure.

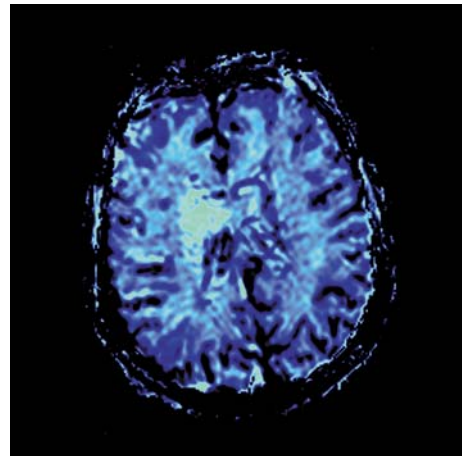
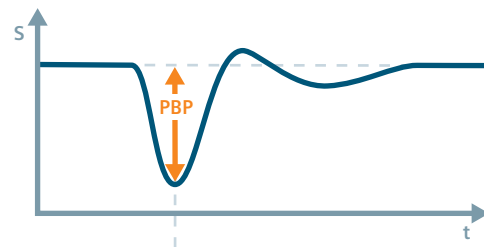
The MTT map depicts the time blood stays in a volume of interest before being replaced by fresh blood.

Reduced perfusion

The global bolus plot provides only a general description 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 **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 is shown hyperintense.

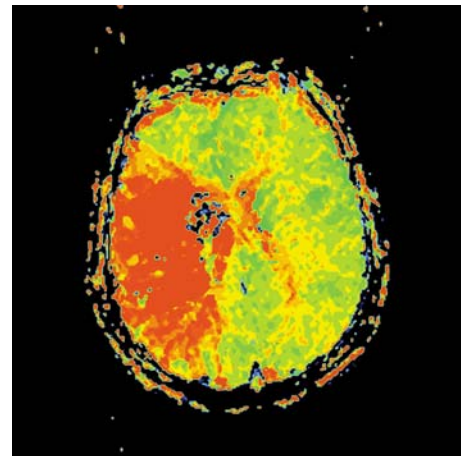
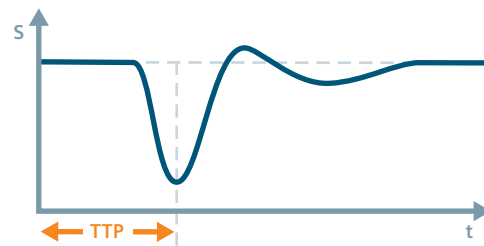


PBP map

Delayed perfusion

The **time to peak** (TTP) is the time interval to the bolus peak. The TTP map shows the regional distribution of the time needed to the minimum perfusion signal, either in gray values or color encoded.

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



TTP map in color, the red pixels indicate delayed perfusion



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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 rCBV, rCBF, and the TTP map. The maps show functional information only.



BOLD imaging

Using magnetic properties of blood to analyze brain activity

Blood carries the signal

Which neural areas are participating in a cognitive process, be it perception, thinking, or movement of limbs? When searching for the respective brain region, we are *not* measuring the neural activity directly, instead we are looking for locally increased oxygen concentrations connected with changes in blood circulation (**BOLD** = Blood Oxygenation Level Dependent).

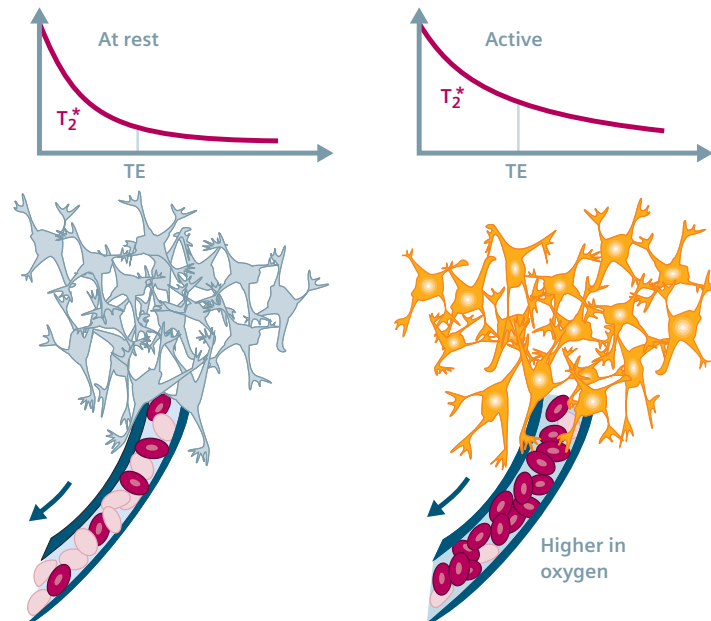
BOLD does *not* measure oxygen consumption. During activity, the cerebral nerve cells' need for oxygen increases. The increase in blood flow ensures that the blood contains more oxyhemoglobin, the carrier of oxygen, after oxygen depletion than during the inactivity of neurons.

Oxygen depletion is overcompensated. An increase in oxygen concentration in a brain region is therefore an indicator for local neural activity.

Susceptibility changes in blood

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

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





Paradigms, t-tests, and mosaics

Correlating brain states at rest with neural activity

Creating t-test images

Let us 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 (*rest*). We follow that with acquisitions where the fingers are being moved for several seconds (*active*). To obtain valid results, we perform this interchange several times, for example, 10 times – that is to say we perform a **paradigm**.

The images taken at rest and those with neural activity are evaluated using 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.

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

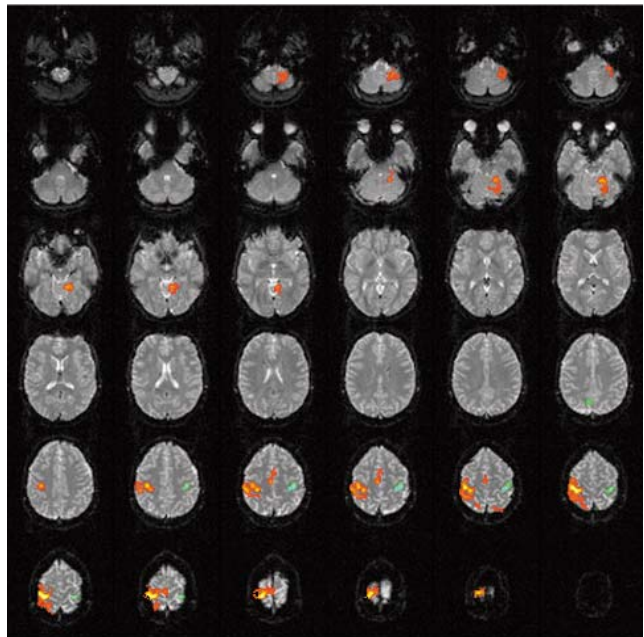
t-test results, 3D superimposed
on anatomical images



Displaying the results in mosaic images

To visualize active brain regions, images of the entire brain are required. Each measurement generates a multitude of images from a series of slices. Thousands of images are measured and computed.

To be able to manage this volume of images, mosaic imaging is used. The software combines and stores the images in a matrix directly after executing the measurement sequence.



Mosaic images

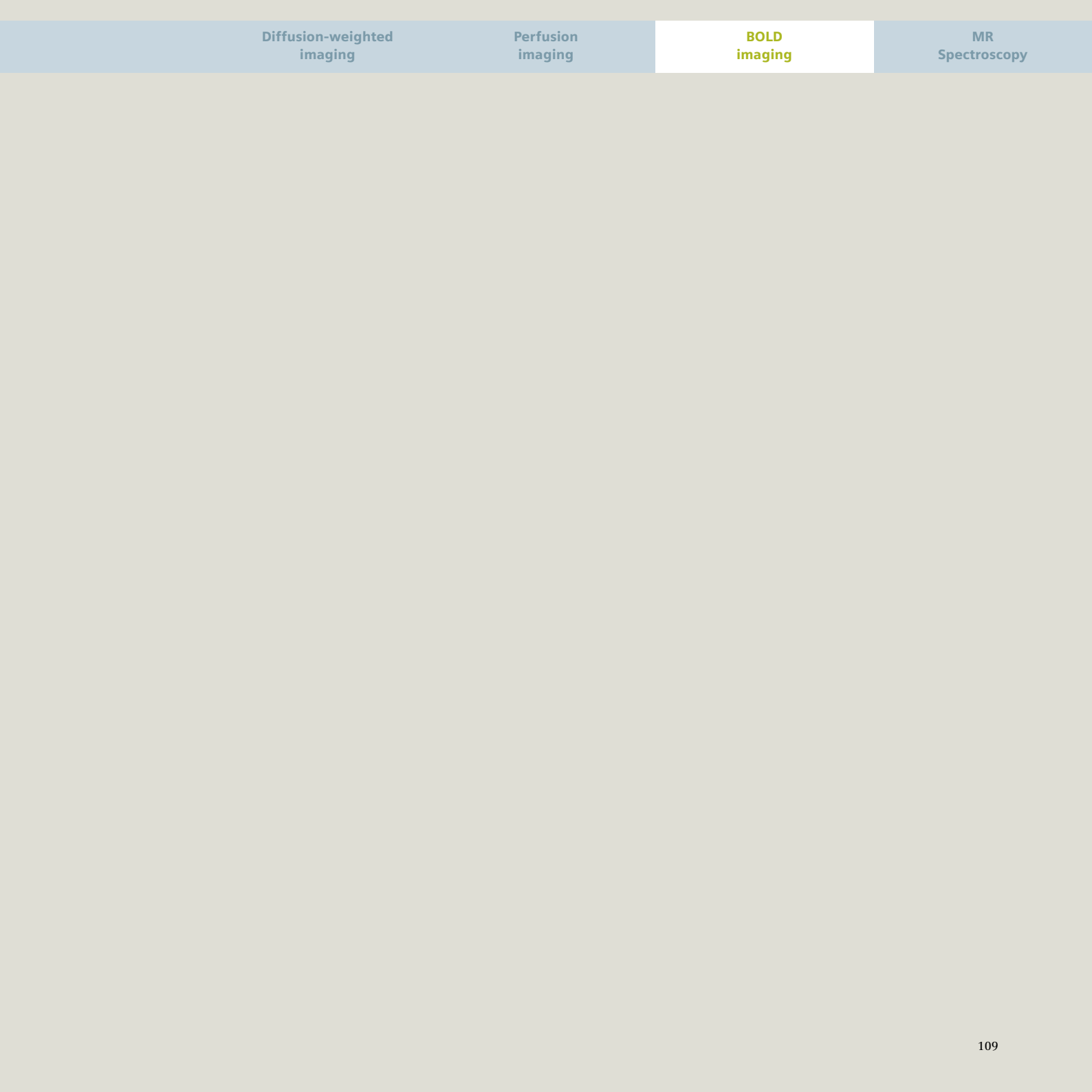


Summary

Neural activity, blood flow and nutrient transport, molecular motion, and biochemical analysis – functional and quantitative imaging provide us with new insights into the human body.

Research in epilepsy, evaluation of vessel closures, or the diagnosis of strokes are just some of the applications addressed by functional MRI.

MR scanners with a field strength of 3 or 7 tesla and higher are able to improve visualization of processes on the molecular and neural level.



Diffusion-weighted
imaging

Perfusion
imaging

**BOLD
imaging**

MR
Spectroscopy

MR spectroscopy

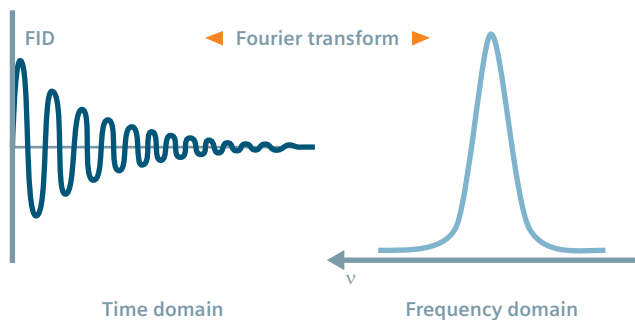
In addition to image generation, MR spectroscopy provides for biochemical assessment

From the FID to the peak

In MR spectroscopy as in MR imaging, the MR signal is measured as a function of time: the free induction decay (FID) is a rapidly decreasing RF oscillation. FID and spin echoes are used in spectroscopic sequences.

By using a single Fourier transform, this oscillation is converted into its frequency components known as the **spectrum**.

This transformation is a conversion of the signal from the time domain into the frequency domain.



If the signal has only one frequency (sine oscillation), the associated spectrum consists of a single **resonance line** only of the associated frequency. Since we are dealing with local magnetic field inhomogeneities and molecular motion resulting in field fluctuations, there is no “sharp” resonance line but a broader **peak** of finite width.

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



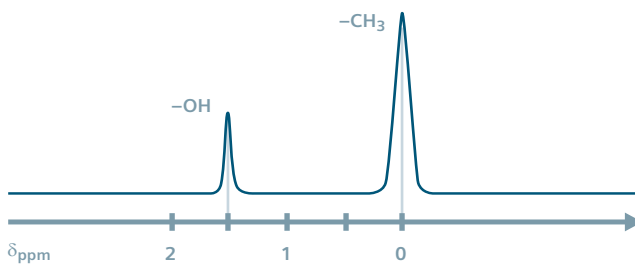
Spectrum, chemical shift, and techniques

An MR spectrum shows the dependence of the signal intensity on the chemical shift (see Chapter 2, *Saturation and Chemical Shift*). The concentration of metabolites contributing to the spectrum can then be quantified. We are able to differentiate molecular components, molecules, and substances.

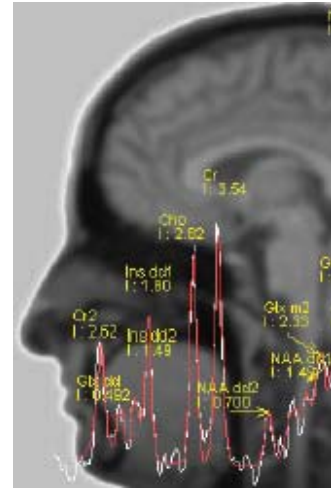
Chemical shift imaging (**CSI**) maps the metabolic information from a volume of interest (VOI) into a spectral matrix.

Single-volume spectroscopy (**SVS**) maps the metabolic information from a volume into a spectrum. SVS is advantageous in case of pathological changes that cannot be spatially limited to a few VOIs.

Example Methanol (CH_3OH): The ratio of the peak areas is 3 : 1. As a result the peaks can be either associated with the hydroxyl group (OH) or with the 3 equivalent hydrogen atoms of the methyl group (CH_3). The degree of chemical shift can be expressed in δ_{ppm} (ppm = parts per million). $\delta_{\text{ppm}} = -1.5$ means the frequency of the OH group has been reduced by 1.5 millionth (this is by 60 Hz at a 40 MHz Larmor frequency).

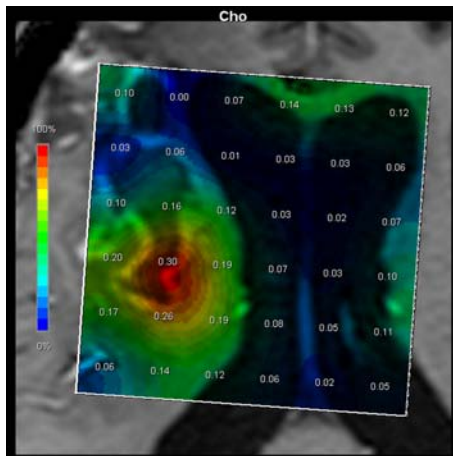


In practical application, the comparison of spectra does not involve absolute peak areas, but rather the relative signal intensities. They are used to compare spectra in healthy tissue with spectra in pathological tissue.

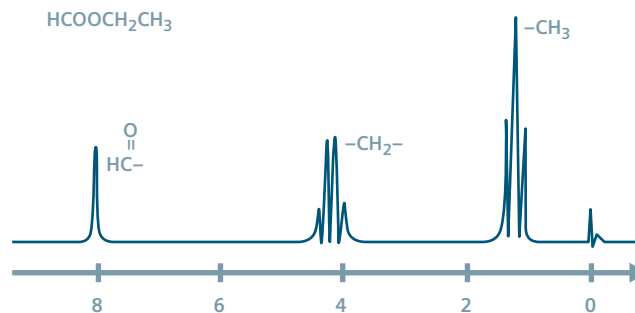


The fine splitting of the resonance lines

Not all nuclei provide simple resonance lines (*singlets*). Some nuclei show a characteristic splitting of the lines, such as triplets or quartets. This is caused by the magnetic interaction of the nuclei (spin-spin coupling).



Choline metabolic map generated from 3D CSI measurement







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 important to learn how to detect and allocate these artifacts.

Patient and tissue-related artifacts

Motion artifacts

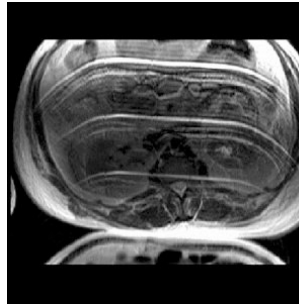
The most pronounced artifacts in the image relate to motion during acquisition: Respiration, heart beat, blood flow, eye movements, swallowing, accidental patient movement.

Ghosting and smearing

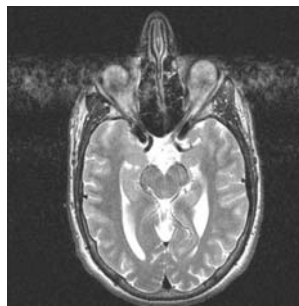
Breathing produces ghosting: the thorax is seen as a locally offset double or multiple structure. Structures rich in signal, for example, subcutaneous fat, further amplify ghosting.

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

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



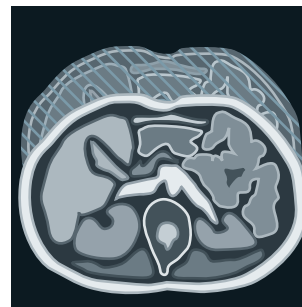
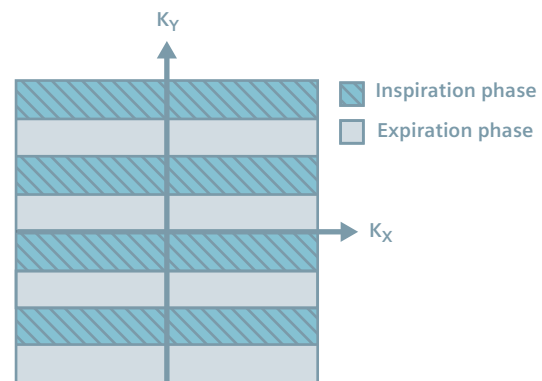
Ghosting in the vicinity of the thorax caused by periodic respiratory movement



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.



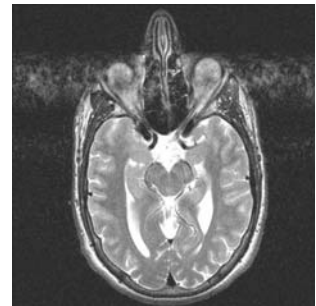
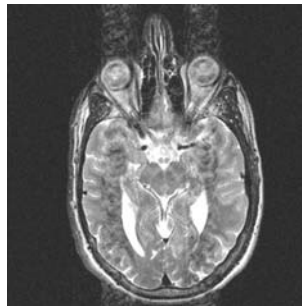


How do we remedy motion artifacts?

Motion artifacts can frequently be relocated to image areas that do not affect interpretation: by swapping phase and frequency encoding (swap).

Other possibilities are:

- fat suppression
- image averaging
- fast imaging, for example, breath-hold exams
- radial techniques (BLADE)
- PACE, etc.



Relocation of motion artifacts by swapping phase and frequency encoding

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

Ghosting is the result of quasi-periodic motion, for example, respiration.

Smearing represents aperiodic structures, for example, eye movement.



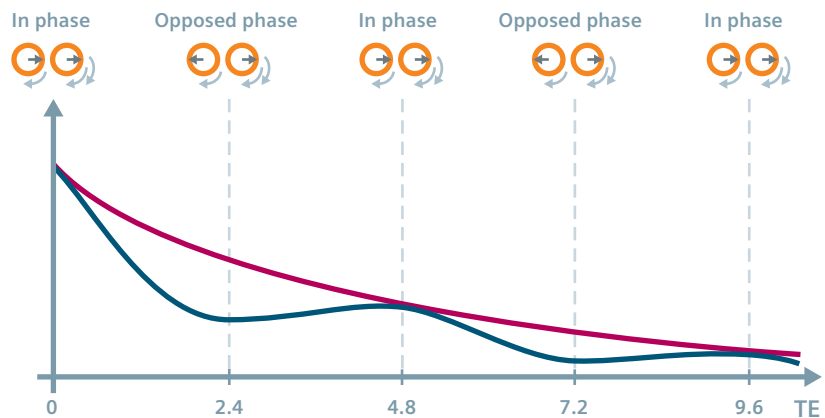
Contour artifact caused by the chemical shift

Phase cancellation in opposed-phase images

Phase cycling

Contour artifacts, also known as 'black boundary' or 'india ink', might appear at the interface of fat and water-containing tissue. The different resonance frequencies of hydrogen protons bound in fat and water (chemical shift) are the source of a possible phase shift affecting the signal.

With a spin-echo sequence, the protons in each voxel precess in phase at the time of readout. With gradient-echo sequences, phase cycling is present: After an excitation pulse, the fat and water spins in a 1.5 tesla magnetic field are alternately 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 and can help in formulating a differential diagnosis.

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

If 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 cancellation” results in contour artifacts that occur across the width of one voxel along the borders of tissue containing fat and water.

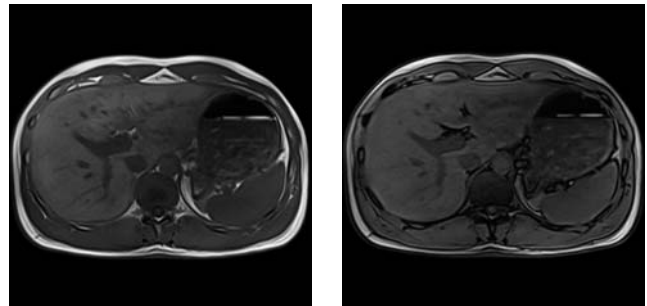


Image reconstruction in phase (left), opposed phase (right)

What do we do about contour artifacts?

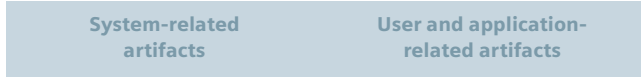
Use an echo time (TE), where fat and water spins are in phase.

0.2 Tesla: TE in phase	36.7 ms, 73.5 ms ...
0.2 Tesla: TE opposed phase	18.4 ms, 55.1 ms ...
1.5 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



**Patient and tissue-
related artifacts**



**System-related
artifacts**

**User and application-
related artifacts**

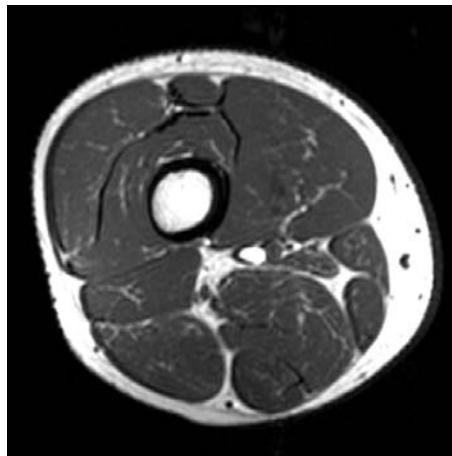


Relief artifacts caused by the chemical shift

Misregistrations in the image

Relief-like structures

Especially susceptible are voxels with direct transitions between tissues with significant differences in fat and water content, for example, vertebrae and intervertebral 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. For chemical shift, see Chapter 2, *Saturation and Chemical Shift*.



Relief artifact

The source of 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 between fat and water, these incorrect encodings result in a bright rim on one side and a dark rim on the opposite side of an anatomical structure in the frequency-encoding (read-out) direction.



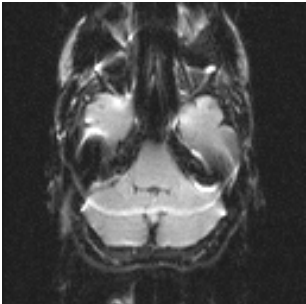
Chemical shift artifacts with EPI

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.

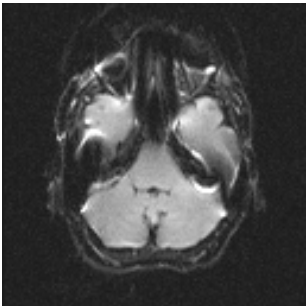
For EPI sequences with their low bandwidths in the phase-encoding direction, the chemical shift of fat and water is seen as a clearly defined artifact **in this direction**

Relief artifacts with 1 Tesla systems

Readout bandwidth	780 Hz
Phase bandwidth	7 Hz
Shift	21 pixels



Shift of fat and water in the head



Artifact elimination in EPI imaging by fat suppression

What do we do with chemical shift 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:

- Using sequences with larger bandwidths
- Swapping the phase and frequency-encoding direction
- Using the STIR sequence
- Fat or water suppression



Distortion artifact (susceptibility)

Altered or shifted signal intensities because of local field inhomogeneities

Susceptibility 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.

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

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, producing 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 heavy signal losses. The local field inhomogeneity is not compensated for.

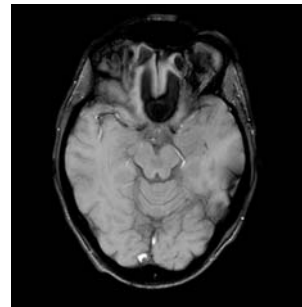
With EPI imaging, the very low bandwidth of the sequences affect additional distortions in the phase-encoding direction.

The higher the field strength of the main magnetic field, the stronger the effect. Conducting and non-ferromagnetic metals will cause signal voids and distortions.

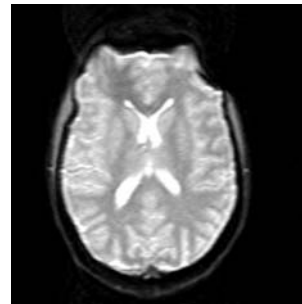
What do we do about susceptibility artifacts?

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

- Using 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 the echo time TE to reduce the period of dephasing.
- Using sequences with higher bandwidths.



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



Susceptibility artifacts in EPI imaging



Susceptibility artifacts caused by ferromagnetic objects on the patient's body

System-related artifacts

Distortion artifacts

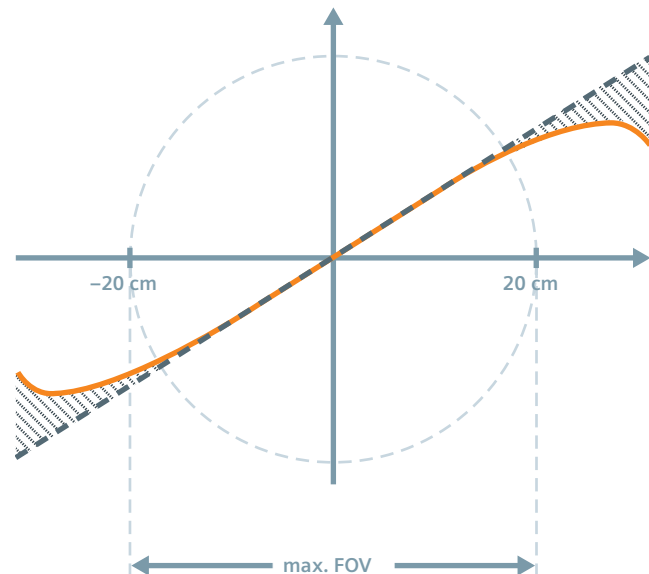
Image distortions at the edge of a large field of view

Sources of distortion

Image distortions are caused by inhomogeneity in the magnetic field, gradient non-linearity, or ferromagnetic materials in proximity to the examination.

Under ideal circumstances, magnetic field gradients rise linearly inside the imaging volume of the magnet. However, in reality, linearity decreases at the edge of the field of view. The size of the field of view that can be acquired by the gradient system, is, for example, 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).



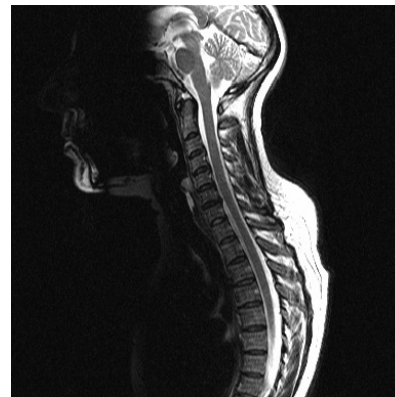
What do we do in case of distortions?

With a so-called “large FOV filter”, taking into account the known gradient field properties, corrections can be performed during image reconstruction.

Disadvantage: The filter prolongs the reconstruction time. In addition, slice positioning in the result images is no longer distinct. For this reason, slice positioning or slice display is not allowed on such filtered images.

Alternatives:

- Table move during scan
- Measure two smaller field of views
- Position the region of interest into the magnet isocenter



Unfiltered original
image with distortion



Distortion correction
using a large FOV filter



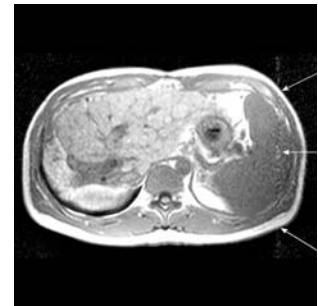
RF interference

Image interference caused by RF irradiation

Sources of RF-interference

RF pulses are transmitted and received during measurement. External RF fields, caused by radios, cellphones, electronic controls, or electrical motors, emit interfering RF signals into the MR systems, which have an adverse effect on image quality.

To protect MR tomographs from external radio-frequency fields, they are installed 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 do we do about RF irradiation?

RF interference is caused by sources outside the examination room. For this reason, always ensure that the door to the room is closed properly.

After structural alterations, newly generated RF fields in the examination room may lead to interference inside the RF shielding, for example, caused by drilling holes for cables. In this case, service engineers need to research carefully the cause and location of possible new sources of interference.

User and application-related artifacts

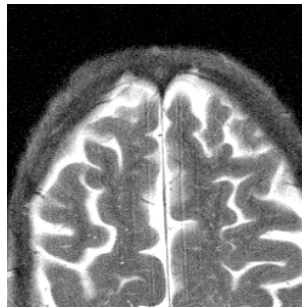
Truncation artifacts (ringing)

Periodic oscillations (stripes or rings) in the image

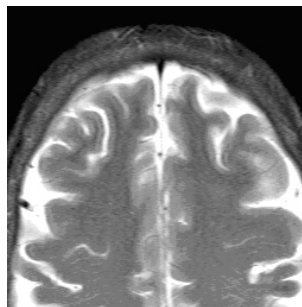
Truncation artifacts and data sampling

Abrupt signal transitions in tissue may lead to artifacts caused by truncation. Periodic oscillations parallel to tissue interfaces are generated. Stripes or rings appear in the image with alternating high and low signal intensity, also called 'edge oscillations.'

An object would be perfectly imaged if an infinite data acquisition window were available. Due to 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 high-contrast tissue interfaces show artifacts caused by truncation.



Edge oscillation without use of a filter



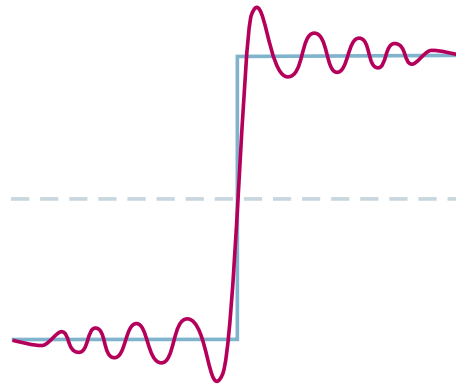
Weak filter application with minimal loss in sharpness

The source of edge oscillation

The edge oscillation artifact is caused by a limited number of sampling points during scanning and subsequent Fourier transform. The Fourier transform is also limited in the number of Fourier elements used. Abrupt signal transitions are simulated by curve approximations.

The Fourier transform approximates a jump in signal intensity at the edge of the object by means of 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 also known as *ringing*.





What do we do about truncation artifacts?

- Use a weak raw-data filter (Hamming filter). The strength of the filter determines the extent of loss in sharpness.
- Increase the size of the measurement matrix.

Wrap-around artifacts (aliasing)

Overfolding by receiving signal from outside the field of view

Sources of wrap-around artifacts

Wrap-arounds are generated when measured tissue is outside the FOV but still within the sensitive volume of the coil. Signals from outside the FOV overlap the image, but on the opposite side.

The tissue excited outside the sensitive volume of the coil contains higher or lower phase and frequency information. Through misinterpretation during Fourier transform (under-sampling of signal), the tissue areas of the opposite side are allocated within the FOV. For this reason, the image area contains both sets of signal information, the original information and the overfolded contribution. 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 field of view



Oversampling prevents wrap-around artifacts during the measurement

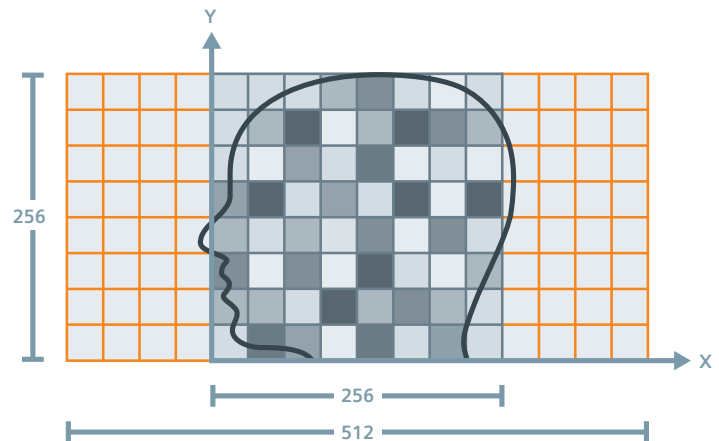
What do we do about wrap-around artifacts?

Wrap-around is successfully avoided by doubling the sampling points (for example, 512 instead of 256) during data sampling/oversampling.

In most cases, this artifact can be observed in the phase-encoding direction. It is recommended to increase the number of sampling points in the phase-encoding direction, however, 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 positioning, the y-axis is set as the phase-encoding direction by default. For coronal slices, this is the x-axis.

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



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Cover image:
Diffusion tensor display from diffusion-weighted MR image of the brain, post-processed using BrainSuite Diffusion Pipeline to show orientation distribution functions (ODFs) in surface view. BrainSuite.org

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