Dominik Weishaupt
Victor D. Köchli
Borut Marincek

How does MRI Work?

An Introduction to the Physics and Function of Magnetic Resonance Imaging

Second Edition

7 Basic Pulse Sequences

Let us once again go through the different steps that make up an *MR pulse sequence.*

— Excitation of the target area
  - Switching on the slice-selection gradient,
  - Delivering the excitation pulse (RF pulse),
  - Switching off the slice-selection gradient.
— Phase encoding
  - Switching on the *phase-encoding gradient* repeatedly, each time with a different strength, to create the desired number of phase shifts across the image.
— Formation of the echo or MR signal
  - Generating an echo, which can be done in two ways (discussed below).
— Collection of the signal
  - Switching on the *frequency-encoding or readout gradient,*
  - *Recording* the echo.

These steps are repeated many times, depending on the desired image quality. A wide variety of sequences are used in medical MR imaging. The most important ones are the spin echo (SE) sequence, the inversion recovery (IR) sequence, and the gradient echo (GRE) sequence, which are the basic MR pulse sequences.

We have already briefly mentioned *echoes* (► Chapter 3) and said that some time must elapse before an MR signal forms after the hydrogen protons have been excited. Now we can explain why this is so:

— Before an MR signal can be collected, the phase-encoding gradient must be switched on for spatial encoding of the signal.
— Some time is also needed to switch off the slice-selection gradient and switch on the frequency-encoding gradient.

— Finally, formation of the echo itself also takes time, which varies with the pulse sequence used.

7.1. Spin Echo (SE) Sequences

Spin echo sequences use a *slice-selective 90° RF pulse* for excitation, after which transverse magnetization decays with T2*, as discussed in ▸ Chapter 2. Dephasing occurs because some spins precess faster than others as a result of the static magnetic field inhomogeneities that are always present. This is why after half of the echo time (TE) has elapsed, a *180° RF pulse* is delivered to reverse or refocus the spins: those spins that were ahead before are now behind and vice versa. However, the spins that are now behind will catch up as they are still exposed to the same field inhomogeneities that caused the phase differences in the first place. Thus, after the second half of the TE interval has passed, all spins meet once again in phase. This is the moment at which the echo forms (▸ Fig. 28). The role of the 180° refocusing pulse in generating the spin echoes can be illustrated by considering a race in which a number of runners start together and, after some time, are given a signal to go back. At the time the signal is given, the fastest runners will have covered the longest distance but also have the longest way back. Assuming that everyone is still running *at their initial speed*, they will all arrive at the starting line together. (The analogy is not quite correct since it is not the direction of precession that is reversed but merely the position of the spins on the precessional path relative to each other. Applied to the example of the race, a magician would have to reverse the order of the runners without their noticing!)

The 180° refocusing pulse then serves to eliminate the effects of static magnetic field inhomogeneities (T2*) but cannot compensate for *variable* field inhomogeneities that underlie spin-spin interaction (T2). Therefore, the magnetization decay that occurs after excitation is slower as it is a function of T2 rather than T2*. Because of this decay, the transverse magnetization component is smaller at the time the echo is collected than immediately after excitation though the decrease in signal is less pronounced than it would be without application of the 180° refocusing pulse. Again, in our analogy, this means that not all runners arrive at the starting line together because they do not always run at a constant speed.

Spin echo sequences are characterized by an excellent image quality precisely because the effects of static field inhomogeneities are eliminated by

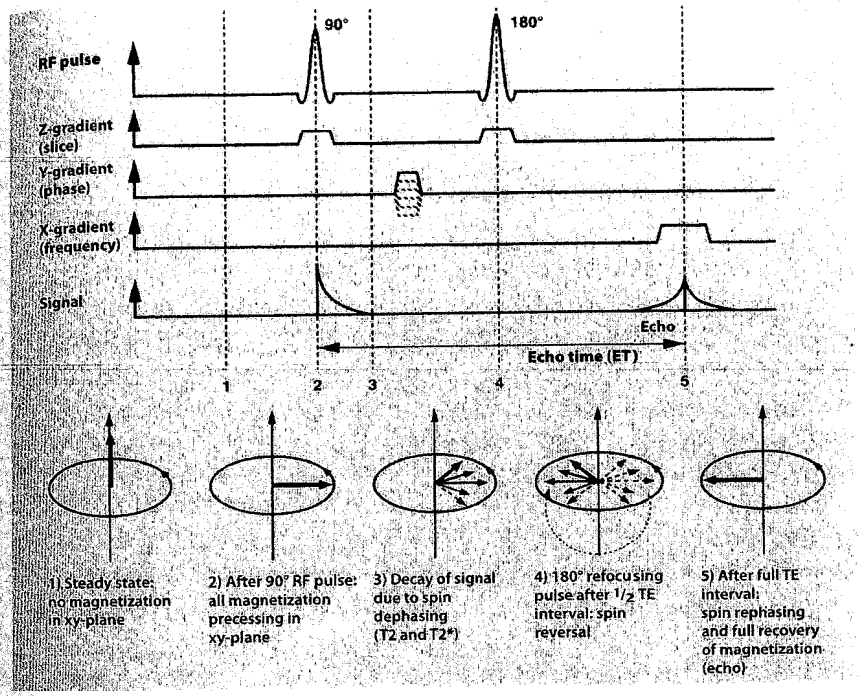

Fig. 28. SE sequence. The excitation pulse always has a flip angle of 90°; the dephased spins are refocused into the spin echo by the 180° pulse. The dashed lines indicate the phase-encoding steps

application of the 180° refocusing pulse. The tradeoff is a fairly long scan time, which makes the sequence highly sensitive to motion artifacts. SE sequences are still used as the standard sequences for acquiring T1-weighted or PD-weighted images. They are preferred for PD imaging because they are less susceptible to motion artifacts compared with FSE sequences.

7.2 Black Blood Effect

The *black blood effect*, or *outflow effect*, refers to a natural high contrast between flowing blood and tissue. It is a specific feature of SE sequences due to the long echo time. Flowing blood appears black because it does not give a signal. This has two reasons:

— All or most of the blood leaves the imaging slice during the long TE and thus the spins are not affected by the 180° refocusing pulse.
— In case of turbulent blood flow, there is additional signal loss due to phase dispersion.

Based on the fact that normal flowing blood is black, we can explain those cases where the outflow effect does not occur:

— If there is *slow blood flow*, excited blood stays in the slice and produces a signal.
— Excited blood may also remain within a slice and become visible if a long segment of a blood vessel lies *within the imaging slice*.
— In case of *thrombosis*, a fresh thrombus will yield a bright signal while an older, organized thrombus appears somewhat darker.

7.3 Multislice Imaging

Conventional imaging with inactive repetition times (TR) between two successive excitation pulses is highly inefficient, especially when using sequences with long scan times and long TRs (e.g. scan time of almost 3 min for acquisition of a T1-weighted SE image with 256 excitations and a TR of 500 msec). The "wait times" or "dead times" can be put to good use by exciting and recording signals from other slices during this period. In this way, 12 slices instead of only one can be acquired in the same time (or even up to 30 slices for T2-weighted sequences with TRs of 2000–4000 msec; ▸ Fig. 29).

A disadvantage of multislice imaging is that, due to imperfect slice profiles or RF pulses, protons outside the selected slice will also be excited. As a result, there will be less longitudinal magnetization and a weaker MR signal.

7.4 Inversion Recovery (IR) Sequences

Inversion recovery (IR) sequences are typically used for *T1-weighted or fat-suppressed imaging* but they can also be used to acquire T2-weighted images.

An IR sequence is an SE sequence with an additional 180° inversion pulse that precedes the usual 90° excitation pulse and 180° rephasing pulse of a conventional SE sequence. The inversion pulse flips longitudinal magnetiza-

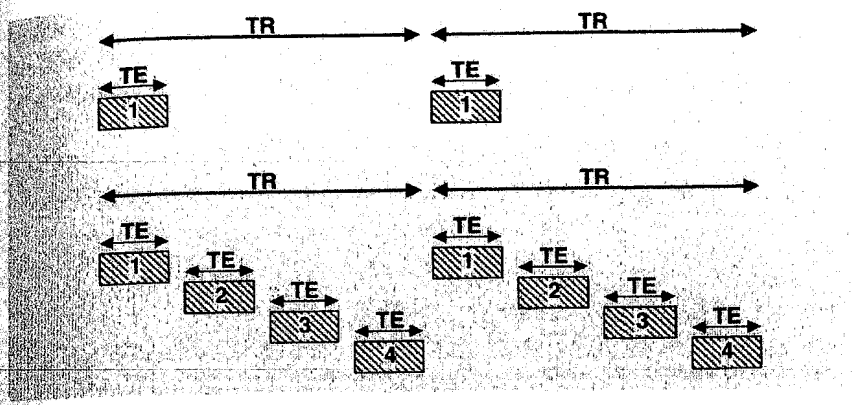

Fig. 29. Multislice imaging (interleaved acquisition). The inactive repetition time, TR, for the first slice is used productively to acquire data from other slices. In the example shown, we thus obtain four slices instead of only one in the same time. (The rectangles represent the different slices)

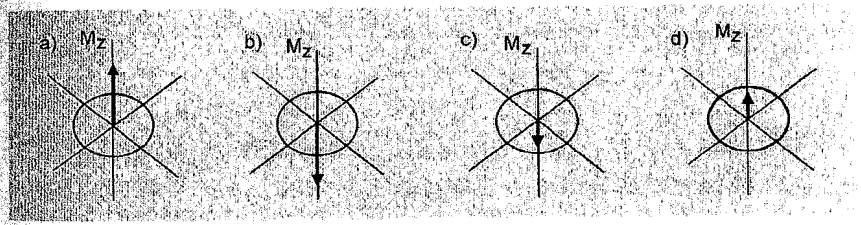

Fig. 30a–c. Inversion recovery sequence with T1 relaxation. Following the 180° inversion pulse **(a)**, the longitudinal magnetization vector points in the opposite direction **(b)**. T1 relaxation takes place from –z to +z **(c, d)**. No signal forms as long as there is no vector component in the transverse plane (the null point of a tissue)

tion from the positive z-direction into the negative z-direction (▸ Fig. 30), which is indicated by the longitudinal magnetization vector now pointing in the opposite direction. As no component of the magnetization vector is in the transverse plane, no signal forms after delivery of the 180° RF pulse. Instead, the inverted longitudinal magnetization vector moves through the transverse plane to return to its original orientation. After some relaxation has occurred, the 90° pulse of the SE sequence is applied. The time between the 180° pulse and the 90° RF pulse is the *inversion time (TI)*.

Image contrast can be manipulated by changing the inversion time. With a short TI and delivery of the 90° excitation pulse immediately after the 180° inversion pulse, all negative longitudinal magnetization is flipped into the

transverse plane. With a longer interval, less longitudinal magnetization is tilted into the transverse plane and a weaker signal is generated. If, however, inversion time is long enough to allow full relaxation, the signal again becomes stronger.

Two IR techniques are widely used in routine clinical applications: the short TI inversion recovery (STIR) sequence and the fluid-attenuated inversion recovery (FLAIR) sequence.

7.5 STIR Sequences

STIR (short TI inversion recovery) sequences are widely used for fat suppression because they reliably eliminate the signal from fat at all magnetic field strengths. A standard STIR sequence inverts the longitudinal magnetization of both fat and water by delivery of the 180° pulse, which is followed by a TI of some hundred milliseconds. To suppress the fat signal, the TI is adjusted such that the 90° RF pulse is emitted exactly at the moment when fat passes through zero. The TI for fat suppression is about 150 msec at a field strength of 1.5 T and about 100 msec at 0.5 T.

7.6 FLAIR Sequences

FLAIR (fluid-attenuated inversion recovery) is an inversion recovery technique that differs from STIR in that very long TI values (typically about 2000 msec) are used. Another difference is that FLAIR sequences are FSE sequences. With such long inversion times, there is nearly complete suppression of the signal from cerebrospinal fluid (CSF) while there is excellent detection of signals from brain tissue, tumors, edema, and fat. FLAIR sequences a very useful for detecting lesions with a poor contrast to surrounding brain tissue.

7.7 Gradient Echo (GRE) Sequences

Gradient echo sequences are also known as *gradient-recalled echo* or *fast-field echo* (FFE) sequences. As suggested by the name, GRE sequences employ *the gradient coils* for producing an echo rather than pairs of RF pulses. This is done by first applying a frequency-encoding gradient with negative

polarity to destroy the phase coherence of the precessing spins (*dephasing*). Subsequently, the gradient is reversed and the spins *rephase* to form a gradient echo (► Fig. 31).

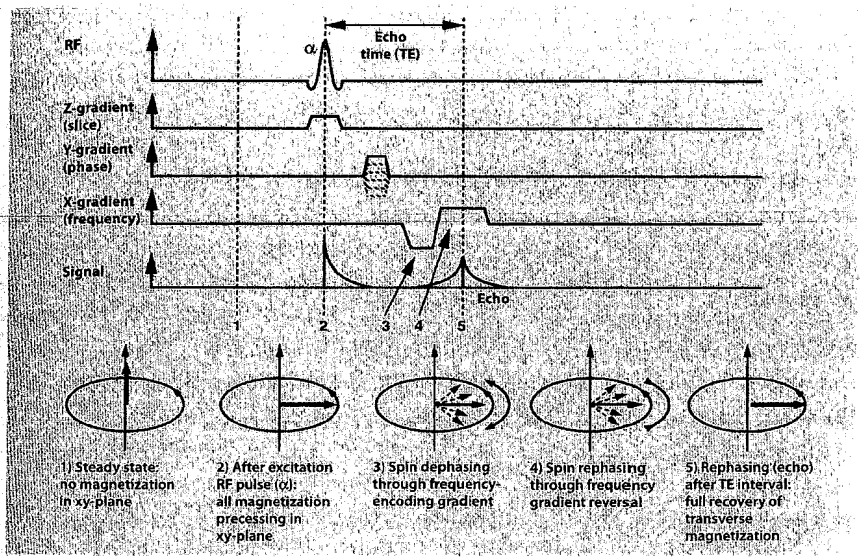

Fig. 31. Gradient echo sequence. For the sake of simplicity, a flip angle α of 90° is assumed here as well

Since no 180° refocusing pulse is needed to generate gradient echoes, very short repetition times (TR) can be achieved. As TR is a major determinant of the overall scan time of a GRE sequence – and of most other sequences – much *faster imaging* is possible compared with SE and IR sequences, which is the most important advantage of GRE imaging. As a result, GRE sequences are less frequently troubled by motion artifacts and are thus preferred whenever a short scan time is desirable. A disadvantage of a short TR is that the time available for T1 relaxation is also short. This may lead to saturation and reduce the SNR when a large flip angle is used. Because no 180° RF pulse is delivered, static field inhomogeneities are not compensated for and the signal decays with T2*. The image contrast resulting from differences in the T2* decay of various tissues is called *T2* contrast*. The T2* contrast of GRE images is affected by TE, which should be as short as possible to achieve optimal T1 weighting (to minimize T2* contrast

and to reduce susceptibility effects). Conversely, a longer TE is selected to accentuate T2* contrast. T1 effects are minimized by simultaneously using a long TR. T2*-weighted images are useful to detect calcifications or deposits of blood products in tissues with a very short T2 such as connective tissues. GRE sequences are also used in conjunction with the administration of iron oxide-based contrast media (► Chapter 12).

One problem, however, needs to be briefly mentioned. Since some GRE sequences are very fast and use very short repetition times, it is highly likely that part of the signal will be "left over" from cycle to cycle. This signal must be destroyed when T1-weighted images are acquired. The purposeful destruction of the residual MR signal is called *spoiling* and is accomplished by turning on the slice-select gradient an additional time to dephase the spins before the next RF pulse is applied. Spoiled GRE sequences are widely used in the clinical setting and are available from all manufacturers of MR scanners.

Popular spoiled GRE sequences are SPGR (spoiled gradient echo) and FLASH (fast low angle shot). The contrast in spoiled GRE sequences can be manipulated as follows:

— T1 weighting increases as TR decreases;
— T1 weighting increases with the flip angle;
— T2* weighting increases with TE.

Proton density-weighted images are generated with a fairly long TR (100–400 msec), a low flip angle (≤20°), and a short TE (5–10 msec). T2*-weighted images result when a long TR (20–500 msec) and long TE (2–50 msec) are used. T1 weighting is achieved by a short TR (20–80 ms), short TE (5–10 msec), and a flip angle of 30–50°.

Spoiled GRE sequences can be acquired in the 2D or 3D mode. The 3D spoiled GRE technique enables volumetric thin-slice imaging without interslice gaps and allows for multiplanar reformatting.

A special type of GRE sequence used for routine MR imaging is the steady-state free precession (SSFP) sequence. SSFP is an unspoiled sequence in that part of the phase coherence of transverse magnetization is preserved from one TR interval to the next. This means that the transverse magnetization generated with a single RF pulse contributes to the formation of several echoes. Various acronyms are used by different manufacturers to designate SSFP sequences such as GRASS (gradient-recalled acquisition in the steady state) or FISP (fast imaging with steady-state precession). Further developments of the SSFP technique are FIESTA (fast imaging employing steady-state acquisition), balanced FFE (fast-field echo), and true FISP. FIESTA and true FISP are T2-weighted GRE sequences whose image contrast

is determined by the T2/T1 ratio. Blood has a high T2/T1 ratio and therefore appears bright on SSFP images. Another advantage of SSFP is that it is not very prone to flowing blood. SSFP sequences are characterized by very short scan times and are thus well suited for vascular imaging and real-time imaging of moving organs such as the heart (► Chapter 11.6).

7.8 Multiecho Sequences

Several echoes can be generated in a single cycle with both SE and GRE sequences: additional spin echoes are produced by applying extra 180° refocusing RF pulses while multiple gradient echoes are generated by repeat reversal of the frequency-encoding gradient. Multiecho techniques are employed for two reasons:

— The generation of multiple echoes enables acquisition of a sequence with *several measurements that differ in their echo times and T2 weightings*. For instance, a repetition time of 2000 msec with echo times of 20 msec for the first and 80 msec for the second echo allows acquisition of a proton density-weighted image (20 msec) and a T2-weighted image (80 msec) with a single measurement. The multiecho technique is routinely used in the clinical setting (► Fig. 32).
— The multiecho technique *accelerates data acquisition* and can be used for ultrafast imaging (► Chapter 8).

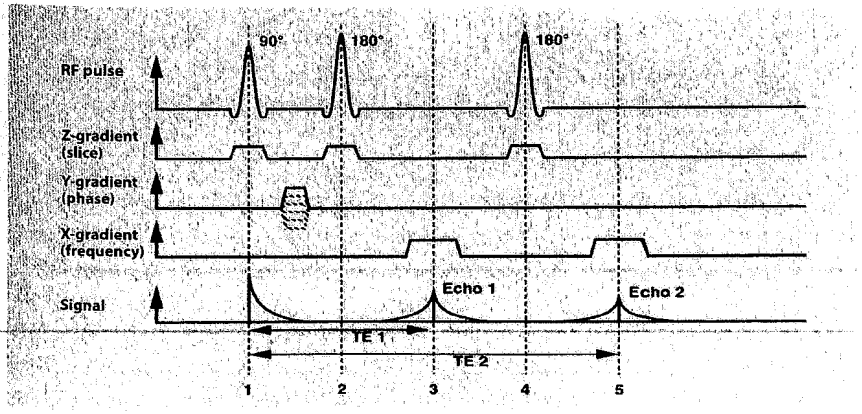

Fig. 32. Multiecho SE sequence. A second 180° refocusing RF pulse (*4*) is applied to generate a second echo (*5*), resulting in an image with heavier T2 weighting due to the longer TE. The second 180° pulse is delivered exactly midway between the first (*3*) and the second (*5*) echo

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8 Fast Pulse Sequences

There are several reasons why it is desirable to speed up scanning.
- A fast sequence allows one to perform dynamic studies, e.g. to track a contrast medium bolus.
- Shorter acquisition is less prone to motion artifacts, which is especially important in uncooperative patients.
- A sequence that is fast enough can be acquired during breath-hold and thus yields images without respiratory artifacts.

Various techniques are available to shorten scan time:
- Use of state-of-the-art gradient and RF systems to full capacity and more effective timing of conventional sequences ([ultra-]fast GRE).
- Sampling of multiple echoes with different phase encodings (FSE, echo planar imaging).
- Incomplete filling of k-space (fractional echo imaging, partial Fourier imaging, rectangular field of view).

8.1 Fast or Turbo Spin Echo Sequences

Fast spin echo (FSE) sequences (also called turbo spin echo (TSE) sequences by some manufacturers) are modified SE sequences with considerably shorter scan times. This is accomplished by delivering several 180° refocusing RF pulses during each TR interval and briefly switching on the phase-encoding gradient between echoes. In this way, optimal use is made of the TR interval by sampling several echoes *with different phase encodings* after each excitation pulse (► Fig. 33). The series of spin echoes thus generated is called an *echo train* and the number of echoes sampled is the *echo train length (ETL)*. The imaging time of an FSE sequence is calculated as:

Scan time = TR × number of phase-encoding steps × number of signal averages [ETL]
ETL is the *echo train length* and refers to the number of echoes sampled per echo train.

FSE sequences are not only faster but differ from conventional SE techniques in a number of other ways as well.

— FSE sequences have a longer TR in order to deliver as many 180° refocusing RF pulses as possible. The TR of FSE is 4000 msec or greater compared with 2000–2500 msec for SE sequences. With their longer TR, FSE sequences are well suited for the acquisition of T2-weighted images.
— The TE of FSE sequences for T2-weighted images is also longer.

The fact that several echoes can be generated after a single excitation pulse is exploited in conventional imaging to acquire a proton density-weighted (intermediate-weighted) image and a T2-weighted image with the same sequence (► Chapter 7.8). Alternatively, the multiecho technique can be used to acquire faster sequences.

FSE sequences can be used to perform double echo imaging by splitting the echo train. With an echo train length of eight, for example, the first four echoes can be used to generate a proton density-weighted image and the last four echoes to generate a T2-weighted image.

8.2 Single-Shot Fast Spin Echo (SSFSE) Sequences

Single-shot fast spin echo (SSFSE) and half-Fourier acquisition single-shot fast spin echo (HASTE) are alternative names for a very fast MR technique with scan times of 1 sec or less. The technique is based on incomplete k-space filling (fractional echo and partial Fourier imaging). "Single-shot" indicates that half of the k-space lines are filled after only one RF excitation pulse. The speed of acquisition reduces motion artifacts to a minimum. Because of the long echo times, SSFSE or HASTE images selectively depict tissues with long TEs, i.e. compartments containing free liquid, whereas tissues with short or medium-length TEs are not shown. For this reason, the SSFSE or HASTE technique is used for MR myelography, MR urography, and MR cholangiopancreatography (MRCP).

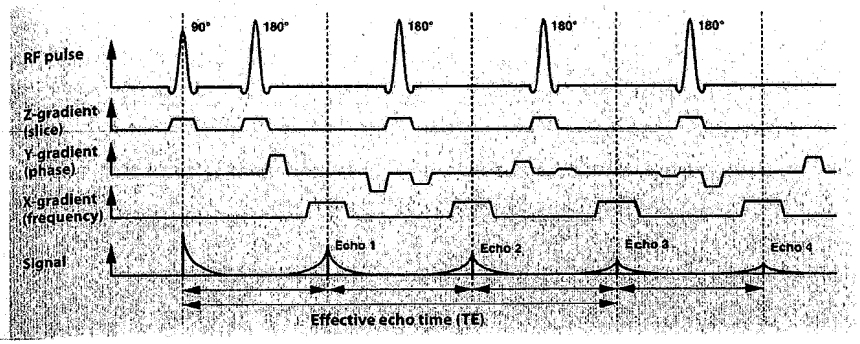

Fig. 33. Fast spin echo sequence. Four 180° refocusing RF pulses are applied to create four echoes (echo train). Since, in contrast to the multiecho technique, the phase-encoding gradient is switched on before each echo, the four echoes obtained after a single excitation pulse have different phase encodings. In the example shown, T2 contrast is determined principally by the third echo (effective TE, ► Chapter 8.9)

8.3 Fast or Turbo Inversion Recovery (Fast STIR) Sequences

Modifying the echo trains of an IR sequence is especially effective because the extremely long TRs allow for full T1 relaxation to occur. Fast or turbo inversion recovery (fast STIR) sequences have the same inversion time as conventional STIR sequences and also use an initial 180° inversion pulse but sample all echoes of an echo train with different phase encodings.

8.4 Fast Gradient Echo (GRE) Sequences

Fast gradient echo (GRE) sequences (also known as *turbo gradient echo* or *ultrafast gradient echo sequences*) used in conjunction with state-of-the-art gradient systems (active shielding) achieve echo times below 1 msec with repetition times of 5 msec or less. Fast GRE is basically a conventional GRE sequence that is run faster and uses some mathematical tricks, primarily incomplete filling of k-space (fractional echo and partial Fourier imaging, ► Chapter 5.3). Fast GRE sequences yield an excellent image quality although a slice can be acquired in only a few seconds (typically 2–3 sec). Such sequences are highly suitable for dynamic imaging, for example, to track the inflow of a contrast medium bolus. Moreover, fast GRE techniques

are used for imaging body regions where motion artifacts must be eliminated such as the chest (respiratory motion) and the abdomen (peristalsis).

Fast spoiled GRE techniques employ a smaller flip angle, typically less than 45°, for optimal T1 weighting. This improves SNR since there is less time for T1 relaxation when TR is short (saturation, ▶ Chapter 3).

8.5 Echo Planar Imaging (EPI) Sequence

Echo planar imaging (EPI) enables ultrafast data acquisition, making it an excellent candidate for dynamic and functional MR imaging. This method requires strong and rapidly switched frequency-encoding gradients. An echo train consisting of up to 128 echoes can be acquired (▶ Fig. 34). In this way, it is possible to obtain an image with a resolution of 256×128 after a single excitation pulse (single shot) in 70 msec, which corresponds to 16 images per second! However, EPI still has to tackle a couple of problems, which have so far precluded its routine clinical use. These are:

— As a GRE technique, EPI cannot compensate for field inhomogeneities and the signal decays with T2*.
— The rapidly switched gradients induce field inhomogeneities that accumulate over time, causing geometrical distortions of the MR image.
— Due to rapid T2* decay of the signal, there is only little time for echo collection. To perform an adequate number of measurements in the short interval available, a very strong and fast gradient is needed. The speed of gradient switching is limited by the electrical inertia of the gradient coils and by the risk of damage to the person being imaged as a result of nerve stimulation associated with rapidly changing magnetic fields. Moreover, rapid gradient switching is so noisy that patients need ear protection!
— Image contrast is often rather poor since a single-shot acquisition involves no repetition and hence there is no T1 effect. Contrast can be improved by applying a presaturation pulse but only at the expense of the signal-to-noise ratio, which is already poor.

8.6 Hybrid Sequences

Hybrid techniques generate and record a series of alternating SEs and GREs. GRASE (gradient and spin echo) and spiral imaging are hybrid techniques.

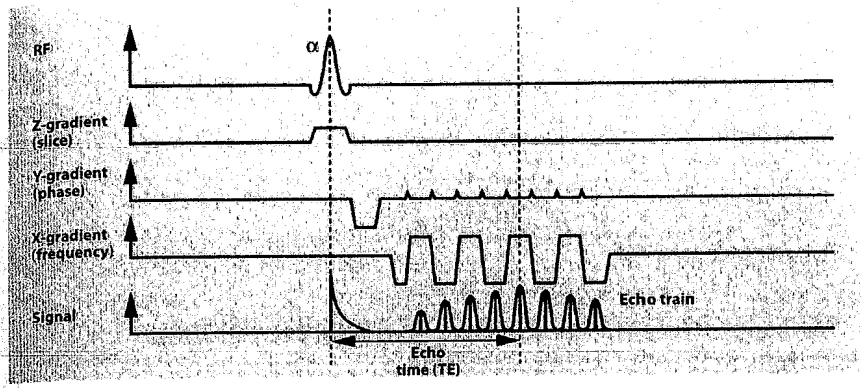

Fig. 34. Echo planar imaging (EPI). As with the FSE technique, several echoes (eight in the example shown) are generated with different phase encodings. In contrast to FSE, the echoes are not generated with a 180° RF pulse but with the frequency-encoding gradient – as in a GRE sequence. This technique requires powerful amplifiers since the frequency-encoding gradient must be reversed very rapidly. The peaks of the phase-encoding gradient are called "blips"

8.7 Gradient and Spin Echo (GRASE) Sequence

A gradient and spin echo (GRASE) sequence is a combination of FSE and EPI. A series of 180° RF pulses is applied to generate several spin echoes (as in FSE). In addition, several GREs are produced for each SE by rapidly switching the readout gradient polarity. This makes the GRASE technique even faster than FSE without impairing image quality as the signal decays with T2 rather than with T2*. The contrast achieved is the same as that obtained with conventional SE sequences.

8.8 Spiral Sequences

Spiral sequences derive their name from the fact that k-space is filled using a spiral trajectory. Spiral imaging is performed with a GRE sequence combined with two oscillating gradients. It is a promising approach, especially for real-time imaging of the heart.

8.9 Echo Time and T2 Contrast in Fast Sequences

In conventional SE and GRE imaging, only one echo is formed after each excitation. As a result, all echoes sampled for an image have the same echo time and thus the same T2 weighting. The T2 weighting of an image generated in this way is well defined.

In contrast, fast SE and EPI sequences generate several echoes with *different T2 weightings*, all of which contribute to the contrast of the resulting image. This is why one of the echoes is selected to mainly determine T2 contrast (in ▶ Fig. 33 the third of four echoes). Its echo time is called *effective echo time (effective TE)*. However, we must be aware that the other TEs also contribute to the T2 contrast.

Technically, the echo is selected by recording it in such a way that it fills the center of k-space (▶ Chapter 4.2), which contains the data that most strongly affect image contrast.

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9 Fat Suppression Techniques

Several techniques are employed in clinical MR imaging to reduce (suppress) the signal from fat.

- Chemical shift imaging based on the time-dependent phase shifts between water and fat
- Frequency-selective fat saturation (fat sat pulse)
- T1-dependent fat suppression (STIR)
- Spectral presaturation with inversion recovery (SPIR)

9.1 Chemical Shift Imaging

As already mentioned, the same atomic nucleus differs slightly in its resonant frequency when bound in different molecules or at different molecular sites. This type of resonant frequency difference is known as *chemical shift*. The chemical shift can be given in Hertz (Hz), which is proportional to the strength of an external magnetic field to which the protons are exposed, or as "parts per million" (ppm), a unit which is independent of the magnetic field strength.

The chemical shift most important in clinical imaging is that between protons in fat and water. The resonant frequency of fat protons bound in long-chained fatty acids (e.g. triglycerides) and water protons differs by 3.5 ppm, which, at a field strength of 1.5 T, causes fat to precess 225 Hz slower than water (▶ Fig. 35). If the water and fat protons are in the same voxel, the precessional frequency difference will become apparent as a phase difference after magnetization has been tilted into the xy-plane and transverse relaxation has occurred. Over time, fat and water protons fall alternately *in* and *out of phase* with each other. They are said to be in *opposed phase* when their phase difference is 180°. At 1.5 T fat and water protons will be

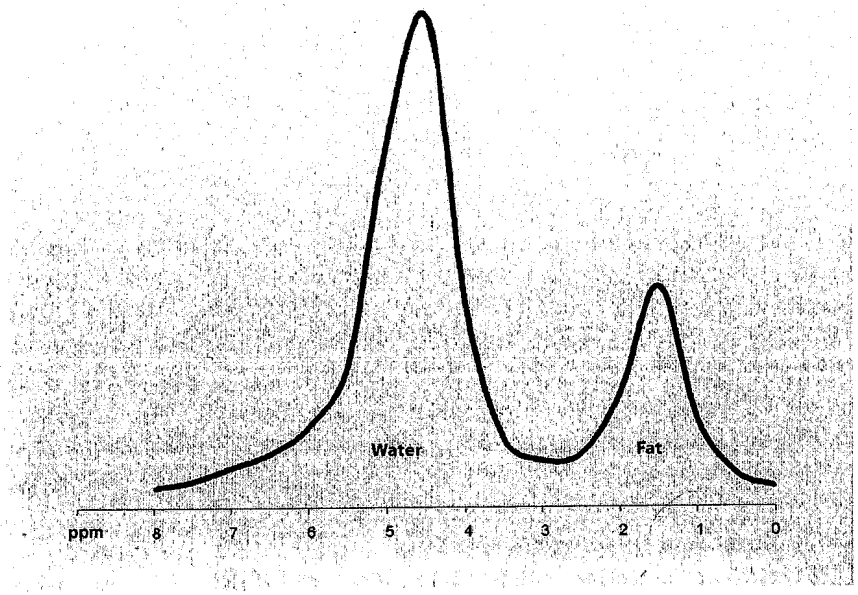

Fig. 35. Chemical shift between fat and water. The resonant frequencies of fat and water protons are separated by approximately 3.5 ppm, which translates into a difference of 225 Hz at 1.5 T

180° out of phase 2.2 msec after excitation and in phase again after 4.4 msec. After another 2.2 msec they will again be out of phase and so on. In clinical MR imaging, these time-dependent phase shifts between the two protons are exploited to suppress the fat (or water) signal selectively. In an image acquired under in-phase conditions, the transverse magnetization components of water and fat protons which are in the same voxel add together and produce a strong signal, while in an out-of-phase image either water or fat alone contributes to the signal (► Fig. 36). The differences in signal intensities between in-phase and opposed-phase images can help differentiate benign and malignant lesions in clinical MR imaging. If an organ lesion contains fat, this will cause a decrease in intralesional signal intensity on the opposed-phase image compared with the in-phase image. This technique is known as *chemical shift imaging* and, for example, has a role in the MR evaluation of adrenal tumors, where the presence of fat is an important criterion for lesion characterization.

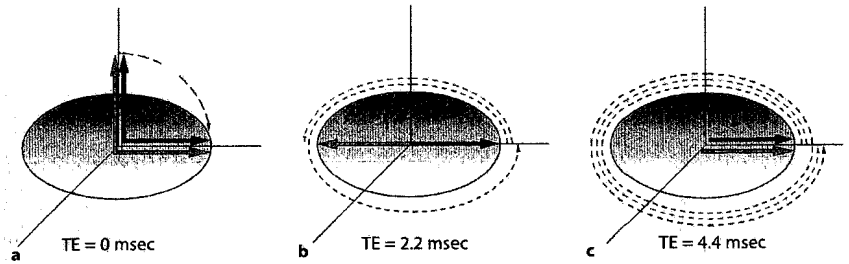

Fig. 36a-c. Phase differences between fat (gray arrow) and water (black arrow) as a function of echo time (TE). At an external magnetic field strength of 1.5 T, the transverse magnetization vectors of fat and water point in opposite directions at TE = 2.2 msec (**b**), resulting in a weak MR signal. (**c**) At TE = 4.4 msec, water and fat are back in phase and both contribute to the MR signal

A technique of chemical shift imaging for the selective suppression of the signals from either fat or water was proposed by *Dixon*. In this method two sets of images are acquired, one with fat and water signals in phase and the other with fat and water signals out of phase. The signal intensities of the two images obtained with this method (image 1 and image 2) can be described as:

— Image 1 = water plus fat
— Image 2 = water minus fat

By adding image 1 and image 2, a pure water image (water plus water) is reconstructed while subtracting image 2 from image 1 generates a pure fat image.

9.2 Frequency-Selective Fat Saturation

Because water and fat have different resonance frequencies, it is possible to selectively saturate the spectral peak of either water or fat by applying a frequency-selective RF pulse before imaging. "True" saturation methods deliver the RF pulse after calibration has been performed to exactly determine the spectral peak of fat. These methods are frequently used in MR spectroscopy but not in routine clinical MR imaging, where fat suppression is generally accomplished by means of a *spoiling* technique. A fat sat pulse is a short frequency-selective 90° RF pulse that is applied to rotate the fat magnetization into the transverse plane. While in the transverse plane, the

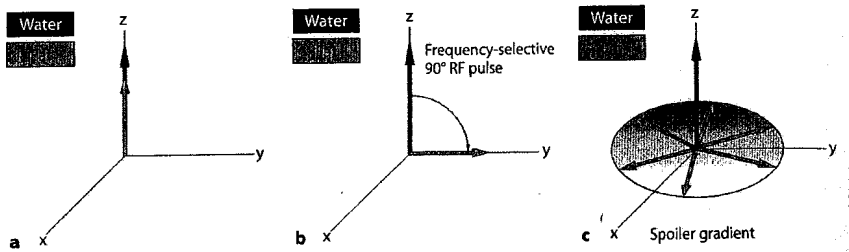

Fig. 37. Frequency-selective fat suppression. A frequency-selective 90° RF pulse is applied to rotate the fat magnetization vector into the transverse plane **(a, b)**. The fat spins begin to dephase, which is accelerated by applying a spoiler gradient. Thus, only the longitudinal magnetization of water is available for subsequent excitation **(c)**

fat magnetization is dephased by application of a spoiler gradient, leaving only the longitudinal magnetization of water for excitation during the next cycle (▸ Figs. 36 and 37).

Frequency-selective fat suppression techniques are typically used on high-field scanners while STIR sequences are preferred on low-field scanners.

9.3 Short TI Inversion Recovery (STIR)

STIR sequences provide reliable fat suppression at all field strengths. They are mainly used for fat suppression on low-field scanners and in all other instances where adequate fat suppression cannot be achieved by means of frequency-selective techniques. The principal function of the STIR sequence is described in ▸ Chapter 7.5.

9.4 Spectral Presaturation with Inversion Recovery (SPIR)

SPIR is similar to STIR in that it is an inversion technique for fat suppression. However, while the STIR sequence uses an initial 180° saturation pulse, the SPIR technique employs an initial inverting pulse that is made frequency-selective and only inverts fat magnetization. Note that SPIR is not a pulse

sequence but merely an additional module that can be applied prior to other pulse sequences. The SPIR module is typically used to obtain fat-suppressed images in conjunction with a T1-weighted sequence.

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of sampled k-space lines is reduced. It can take on any whole-number or fractional value between 1.0 (no acceleration) and about 3.0 to 4.0. Even faster data acquisition is possible with 3D techniques that achieve further acceleration by virtue of their two phase-encoding directions.

Commercially available parallel imaging software is marketed as SENSE, IPAT, ASSET, or SPEEDER. The faster scan time achieved with these tools is of use in a wide range of practical applications. In the clinical setting, a reduction of scan time is especially attractive for imaging protocols with very long sequences or imaging during breath-hold. Short scan times are also beneficial in dynamic MR studies such as evaluation of contrast medium passage or cardiac motion. Alternatively, parallel imaging techniques can be employed to improve spatial resolution or to acquire more slices without unduly increasing scan time.

Finally, parallel imaging can help reduce artifacts. When sequences with long acquisition times are used, shorter readout trains can reduce undesired effects that interfere with image quality. This applies especially to echo planar imaging (EPI), which is frequently degraded by considerable artifacts caused by field inhomogeneities due to variable susceptibility, movement, and flow. Moreover, the extremely rapid gradient reversal necessary in EPI is associated with a very high noise level. Parallel imaging is less noisy because the gradient reversal rate is reduced by shortening the readout train while the overall scan time remains the same.

Whenever one considers applying a parallel imaging technique for any of the reasons outlined, one should also be aware that the sequence used should have some SNR reserve. This is necessary because, with few exceptions, parallel imaging will reduce SNR.

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11 Cardiovascular Imaging

DANIEL NANZ

The cardiovascular system can be examined by MR imaging at different levels.

Vessels are depicted directly (MR angiography, MRA) and can be evaluated for anatomic abnormalities, narrowing, dilatation, or dissection. The advent of new contrast media has dramatically changed vascular MR imaging and has in particular facilitated time-resolved studies. MR images depict not only the blood but also the vessel wall and its diseases.

While blood vessels and capillaries with diameters well below 1 mm are usually not seen directly, perfusion can nevertheless be evaluated using MR techniques which depict tissues with signal intensities that vary with their blood flow. In this way it is possible to directly visualize relative regional differences in organ perfusion.

Effects of *perfusion disturbances* occurring after a stroke can be evaluated on diffusion-weighted MR images obtained within minutes of the onset of symptoms. On such images, the signal intensity reflects the mobility of water molecules at the microscopic level.

In the brain, functional MR imaging provides indirect information on cerebral activity by depicting changes in the *oxygen saturation* of the capillary blood.

MR imaging of the heart presents some specific problems. Notwithstanding, a wide range of clinical questions can be answered by a set of MR images of the *myocardium* or heart muscle obtained with a combination of different sequences.

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11.1 Angiography

Angiographic MR imaging techniques have been optimized to image the blood and surrounding anatomy with different signal intensities. With three of the techniques presented below (time-of-flight, phase-contrast, and black blood MRA), this is accomplished when blood moves faster than surrounding structures. The fourth technique, contrast-enhanced MRA, is different in that a tissue appears bright when its longitudinal relaxation time is shortened to values below 100 msec by administration of a contrast agent. In this way, angiographic contrast agents selectively enhance the blood signal immediately after direct injection into the vascular system.

11.1.1 Bright Blood Imaging

The MRA techniques most widely used in the routine clinical setting depict the blood with a high signal intensity (bright blood imaging). Vessels with positive contrast are more conspicuous and, in electronic postprocessing of MRI data, can be more easily visualized on projections through stacks of images. However, all bright blood techniques are limited by the fact that there is usually no signal from blood when flow is turbulent. Under these conditions, the blood cannot be distinguished from surrounding tissue. Turbulent flow often occurs in important vessel segments such as branchings or vessel segments distal to a stenosis. In general, the only remedy to reduce this effect is to keep the echo time as short as possible.

Angiographic MR techniques can be used to acquire two-dimensional (2D) or three-dimensional (3D) data sets. 2D data can be postprocessed to generate 3D volumes. A general advantage of 3D imaging is that thinner slices can be obtained without interslice gaps, which also improves the signal-to-noise ratio (SNR) in some applications. Moreover, volumetric data sets allow for multiplanar reformation with good resolution. When MRA is performed in the 2D mode, optimal results are achieved with the slices placed perpendicular to the vessel of interest and scanning against the direction of blood flow. This will minimize undesired saturation and partial volume effects (► Fig. 39).

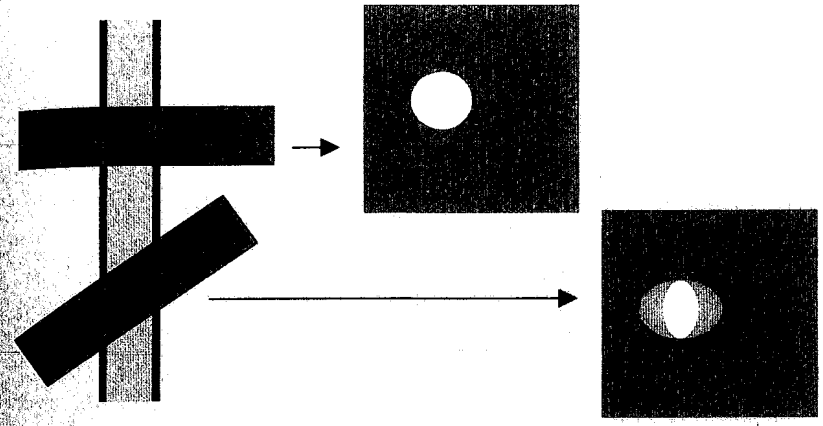

Fig. 39. Partial volume effects occur when the imaged slice is tilted out of the plane perpendicular to the longitudinal vessel axis. The vessel diameter appears smaller on the image from the tilted plane

Time-of-Flight (TOF) MR Angiography

Time-of-flight (TOF) MR angiography depicts blood (rapidly) flowing through the imaging plane with a high signal intensity (bright). TOF angiography is mainly performed in axial orientation for evaluation of the vessels of the head and neck such as the carotid arteries and the circle of Willis. Theoretically, however, TOF MR angiography is an option for vascular imaging throughout the body.

The term "time of flight" is probably adopted from a mass spectrometry technique that separates molecular fragments with different masses on the basis of the different times needed by the fragments to travel through a vacuum tube. In a similar manner, TOF angiography depicts the spins of water molecules that move in the blood through the vessels. A vessel appears bright when there is a continuous supply of "fresh" spins that replace the spins in the imaging plane (inflow effect, ► Fig. 40).

TOF MRA is performed using GRE sequences with short repetition times (30–50 msec). Echo times should be kept as short as possible. The flip angles used range from approximately 20–40° for 3D imaging to 50° or greater for 2D imaging. The spins that rest within the slab without moving are highly saturated by the repeated excitation pulses (► Figs. 11 and 12) and give only

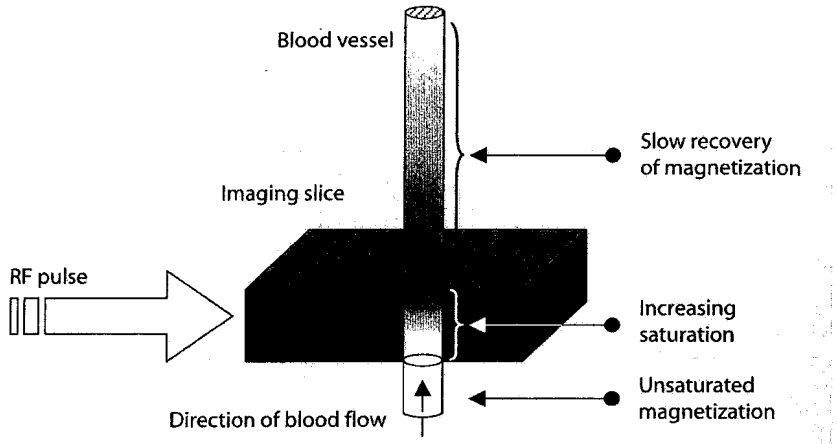

Fig. 40. Principle of TOF angiography. Different shades of gray represent the magnitude of longitudinal magnetization

a very weak signal, making stationary tissue appear dark on the resultant image. In contrast, blood flowing into the imaging plane has not been subjected to these RF pulse. As a result, more longitudinal magnetization is available for subsequent excitation and the inflowing blood appears bright.

If the newly arriving spins do not leave the scan volume within one TR interval, their magnetization will also be saturated by subsequent RF excitation pulses. Their MR signal thus becomes smaller and smaller as they move away from the entry slice. A problem may arise if blood stays in the imaged volume for a long time, for example, if there is slow flow due to vascular pathology (e.g. aneurysm, false lumen, vascular wall sutures, vascular malformations), if vessels take a curved course through the slice, or if a thick slice is acquired (especially in 3D imaging). The increasing signal loss can be mitigated to some extent by gradually increasing the flip angle that is imparted on the spins on their way through the scan volume (tilted optimized non-saturating excitation, TONE). Alternatively, a thick slab can be subdivided (multiple overlapping thin slab acquisition, MOTSA).

Maximum enhancement of flow occurs when thin 2D slices are acquired perpendicular to the direction of flowing blood. This is why 2D MRA techniques may offer advantages in imaging vessels with slow flow such as the portal venous system.

Problems with magnetization saturation may also be encountered when a vessel does not take a straight course but leaves the scan plane and then enters it again. This may lead to very weak signals in distal vessel segments.

The increase in signal induced by inflowing blood is independent of the direction from which the blood enters the imaging plane. For this reason veins are not readily distinguished from arteries in TOF MRA. This problem can be overcome by applying regional presaturation prior to data acquisition. To this end, magnetization is completely saturated either in a slice distal to the imaging slice (arteriography) or proximal to it (venography). Blood flowing into the scan volume from the presaturated slice appears dark (▸ Fig. 41).

The signal from stationary tissue can be suppressed further by saturating the magnetization of the pool of bound protons (▸ Chapter 3.6), which will improve vessel contrast in many instances. Fat suppression is another option to improve contrast.

The presence of moderate concentrations of an MRA contrast medium increases the vessel signal but differentiation of arteries and veins will be more difficult.

The radiologist interpreting TOF MRA images must be aware that the vessel diameter is typically underestimated while a stenosis tends to be overestimated and that contrast may be poor when there is slow blood flow or the vessels do not take a straight course. Also, an unexpectedly bright signal

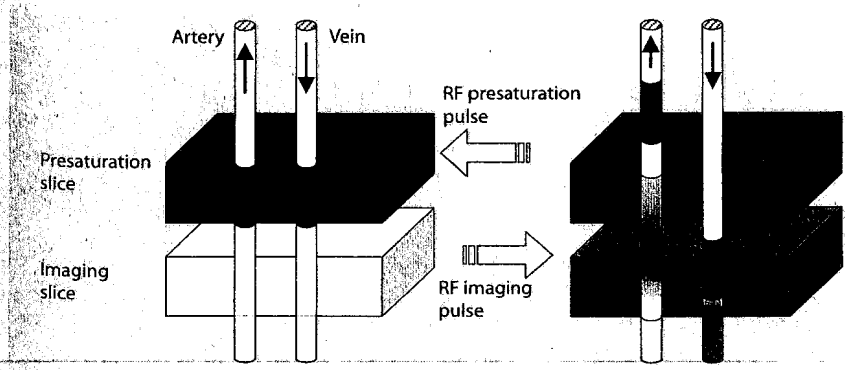

Fig. 41. Differentiation of arteries and veins in TOF angiography. After presaturation of the blood on either side of the imaging slice, the signal intensity of a vessel becomes more dependent on the direction of blood flow

may be seen when relaxation times are shortened by methemoglobin, which may be present in a hematoma or thrombus.

Advantages of TOF angiography are its robustness under routine clinical conditions and efficient data acquisition.

Phase-Contrast Angiography

Phase-contrast (PC) angiography is another bright blood technique and relies on the use of bipolar (flow-encoding) gradients. By selecting the polarity and amplitude of the gradient, the operator can determine the flow direction and the range of flow velocities to which the sequence is sensitive. This technique enables calculation of averaged flow velocities for all voxels imaged.

2D images acquired upstream and downstream of a stenosis, e.g. in a renal artery, can be used to estimate the pressure drop over the stenotic vessel segment. A slice through the stenosis allows one to determine peak flow velocity and the degree of luminal narrowing.

In cardiac imaging, a 2D slice positioned in the ascending aorta just above the aortic valve will provide information on the distribution of outflow velocities over the cross-sectional area of the aorta for "all" (e.g. 20) ECG phases of a cardiac cycle. To this end, a series of 2D phase-contrast angiograms synchronized with the heart rate is acquired at different times during the cardiac cycle (cine phase-contrast imaging). From such a data set, the stroke volume and cardiac output can be estimated. Moreover, an incompetent aortic valve can be diagnosed and the insufficiency quantified by determining the regurgitation volume relative to the stroke volume. Such a velocity profile also provides information on the shearing forces acting on the vessel wall.

3D phase-contrast techniques are mainly used for imaging of intracranial vessels, where excellent results can also be obtained in sagittal orientation.

Phase-contrast MRA sequences are GRE sequences with repetition times in the range of 10 to 20 msec and echo times that should be as short as possible (approximately 5–10 msec). The sequences are made sensitive to flow phenomena by means of a *bipolar gradient* that is applied between the RF excitation pulse and signal readout (► Fig. 42). The flow-encoding gradient pulses induce phase shifts in flowing blood which are proportional to velocity but do not affect the signal from stationary spins (► Fig. 43).

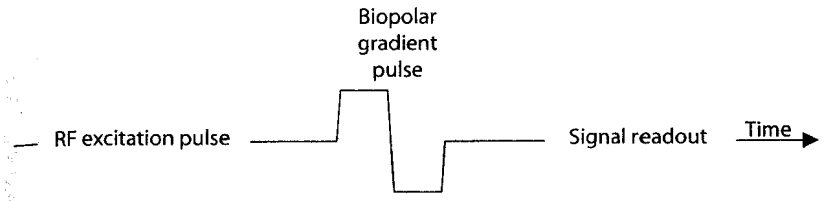

Fig. 42. Diagram of a PC MRA sequence

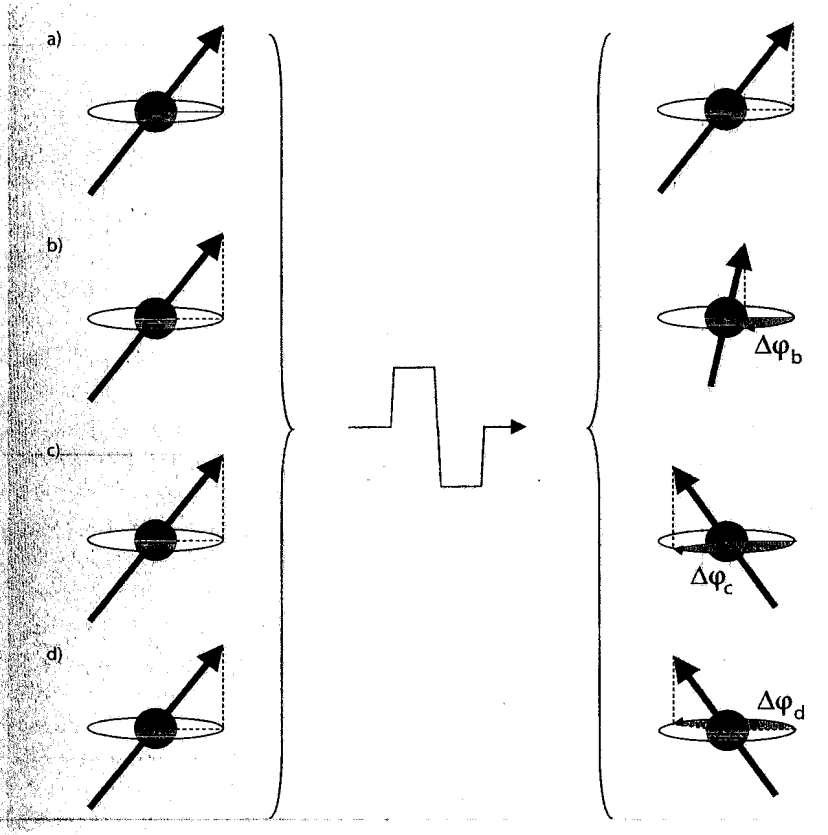

Fig. 43a–d. Delivery of a bipolar gradient pulse and the resulting phase shifts induced in stationary spins (**a**, $\Delta\varphi_a = 0$), spins slowly flowing in the direction of the gradient field (**b**, $\Delta\varphi_b > 0$), spins rapidly flowing in the direction of the gradient field (**c**, $\Delta\varphi_c > \Delta\varphi_b$), and spins rapidly flowing in the opposite direction (**d**, $\Delta\varphi_d = -\Delta\varphi_c$). In a phase-contrast image, the gray scale value of a pixel represents the averaged difference angle, $\Delta\varphi$, measured in the corresponding voxel

The effect of the flow-encoding gradient is negligible for spins that receive both halves of the bipolar pulse at the same site. These spins experience a change in their Larmor frequency as a result of the change in local magnetic field strength and thus precess at a different rate. The second half of the bipolar pulse subjects the stationary spins to a change in magnetic field that is equal in magnitude to that imparted by the first half, only this time the sign is reversed. For stationary spins, the bipolar pulse has therefore no net effect and their phase is the same as if the pulse had never been applied.

The situation is different for spins that move through the field while the bipolar gradient is switched on. Having changed position, these spins are exposed to a different field change by the second half of the pulse. This field change cannot fully compensate for the phase shift imparted by the first half. As a result, there is a persistent phase shift that corresponds in magnitude to the velocity with which the spins move in the direction of the gradient. The phase shift of the spins allows one to calculate blood flow velocity based on the amplitude of the bipolar gradient applied.

The sign of the phase shift is determined by the direction of blood flow relative to the gradient direction. If it is positive for arteries (phase shifts from 0 to +180°) and arteries appear bright on the MR image, it is negative for veins (0 to −180°) and veins appear dark, or vice versa.

Calculation of flow velocities from phase angles between −180° and +180° is straightforward. Problems arise when spins move so fast that their phase shifts exceed +180°. For instance, a phase shift of +200° will be interpreted by the algorithm as a negative phase shift of −160°. As a result, blood flowing near the vessel wall may appear bright while the faster blood in the center of the lumen suddenly becomes quite dark or vice versa. This phenomenon is known as phase wrapping or phase aliasing and can be prevented by properly adjusting the velocity encoding (VENC) parameter. VENC should be chosen to encompass the highest flow velocities likely to be encountered in the vessels of interest. This requires some knowledge of the blood flow velocities in different vascular territories. Arterial flow velocities vary over a wide range from just a few cm/sec to over 200 cm/sec in the ascending aorta. However, one may deliberately choose a low VENC to sensitize the sequence to slow flow. This will also reduce the underestimation of vessel diameters. The VENC parameter adjusts the strength of the bipolar gradient pair and thus the proportionality constant for the phase shift and flow velocity.

The absolute phase of an MR signal is affected by numerous factors and interferences. This is why phase-contrast MRA methods collect data twice

with different flow-encoding gradients. The second data set is acquired with a zero gradient or the polarity of the bipolar gradient reversed (e.g. +/− pulse followed by a −/+ pulse). A systematic error in phase measurement can thus be corrected by subtraction of the two data sets. The tradeoff, however, is a longer overall scan time.

When four data sets are acquired, one without flow encodings and three sets with the bipolar gradients applied along the x-, y-, and z-axes, the three components of the flow velocity vector can be calculated with error correction. In this way, one can generate MR angiograms resembling the images obtained with other angiographic techniques in that flowing blood appears bright. However, a phase-contrast angiogram is superior to other techniques such as TOF because the brightness of the blood exclusively reflects the flow velocity and is not affected by the flow direction.

Moderate amounts of an MRA contrast medium will increase the signal intensity of blood and thus improve SNR.

The acquisition of 3D phase-contrast angiograms with flow encoding in all three spatial directions can be time consuming. Fast flow in large arteries and nearly stagnant blood in an aneurysm or vascular malformation cannot both be adequately depicted with a high sensitivity in a single measurement. Like other techniques, phase-contrast angiography also tends to underestimate vessel diameters and overestimate stenosis.

Advantages of the phase-contrast technique include the quantitative and spatially resolved evaluation of flow velocities and flow directions and the good suppression of the signal from stationary tissue. With proper parameter settings, phase-contrast MR angiography is most suitable for depicting slow flow or flow within the imaging slice. No other MR technique provides the kind of quantitative information that can be derived from the time-resolved velocity and flow profiles obtained with cine phase-contrast angiography for different phases of the cardiac cycle.

Contrast-Enhanced MR Angiography

Blood gives a bright signal on contrast-enhanced MR angiograms if its longitudinal relaxation time is effectively shortened by a suitable contrast medium (► Fig. 44). Contrast-enhanced MR angiography enables rapid acquisition (within seconds) of three-dimensional data sets with a good SNR and a resolution in the millimeter range, thereby allowing imaging of large segments of the vascular system in all regions of the body. Contrast-enhanced

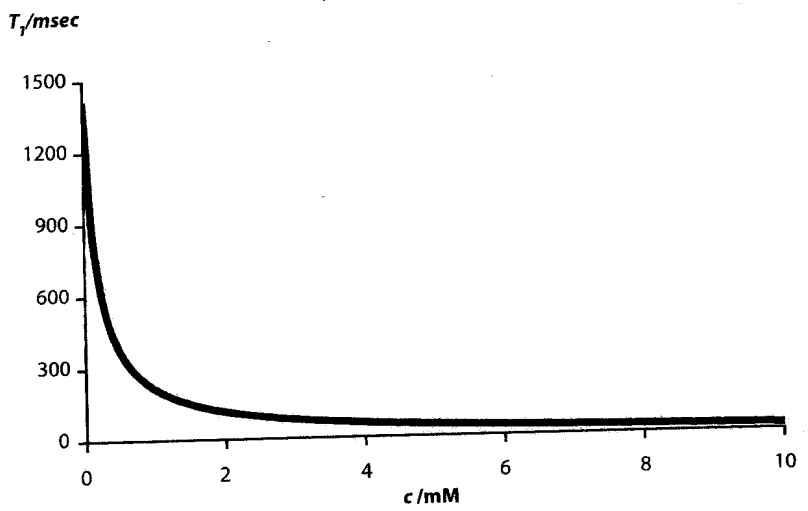

Fig. 44. Shortening of the T1 of water in blood by increasing contrast medium concentrations. Approximate values for a contrast medium with a molar relaxivity of 4 l/(mmol · sec) and a T1 of 1.4 sec without contrast medium

MRA is well established for imaging of major vessels of the trunk and in the periphery but is also used for evaluating the vessels of the head and neck in combination with other techniques.

In general, the contrast agents for MR angiography are injected into a vein in the bend of the elbow. The MRA contrast agents are well-tolerated, gadolinium-based paramagnetic compounds (► Chapter 12) that are administered at doses of 0.05–0.3 millimol gadolinium per kilogram of body weight. In arteriography where a large arterial signal and a small signal from veins are desirable, images must be acquired during the first pass of the contrast medium through the arteries. Arterial enhancement decreases quickly due to the subsequent signal increase in the veins and perfused tissue. Except for the brain, there is rapid diffusion of most contrast media through the capillary walls into the extracellular space. An imaging window of only a few seconds is available from arterial inflow of the contrast medium to its arrival in the veins. This is why proper timing and the duration of the scan are paramount in vascular imaging. Typical scans do not take longer

than 20 seconds. These short scan times allow imaging of the thoracic and abdominal vessels during breath-hold.

When very short scan times are used, a body region can be imaged repeatedly to evaluate the pattern of contrast medium distribution in a time-resolved manner.

Another option is to move the scan plane as the contrast medium bolus advances so as to cover a larger body region with several acquisitions (so-called *multi-station bolus chase*). This is accomplished using an automated table feed technique. The data sets are electronically postprocessed and can be combined to yield a single composite image. Under ideal conditions, this technique allows scanning of the arterial system from head to ankle following a single, optimized contrast medium injection. The well-tolerated MR contrast media available today, however, can be injected repeatedly in a single examination.

Contrast-enhanced MRA is performed with spoiled GRE sequences with very short TRs (approximately 1.7–6 msec) and very short TEs (below 2 msec). Flip angles ranging from approximately 15° to 50° are used. The sequences closely resemble those used for TOF MRA but with a further marked reduction of repetition and echo times. As a result, there is even more efficient suppression of the signal from stationary spins in the scan volume. On the other hand, blood magnetization recovers very quickly when an adequate concentration of a contrast medium is present (on the order of about 5 millimol/liter, depending on the agent administered). In this way, blood gives a strong signal and appears bright despite the repeated RF excitation pulses. Contrast can be further enhanced by combining the technique with fat saturation.

Scan time is a crucial issue and all kinds of tricks are applied to shorten image acquisition. Most techniques available reduce SNR as well. These include:

- Shortening of echo and repetition times through incomplete readout of the data in the frequency-encoding direction (fractional echo imaging, ► Chapter 5.3).
- Reduction of the number of phase-encoding and/or slice-selection steps through incomplete readout of the phase-encoding and slice-select data (partial Fourier imaging, ► Chapter 5.3). The unacquired data is either supplemented on the basis of the conjugate symmetry of k-space or interpolated by applying intelligent algorithms.
- Reduction of the minimum echo and repetition times through a broader receiver bandwidth.

— Parallel imaging (► Chapter 10) with use of suitable receiver coil arrays allows one to further reduce the phase-encoding and/or slice-selection steps or to acquire images with an improved resolution in the same scan time

Moreover, special techniques of k-space ordering can be employed to acquire arterial data during optimal contrast enhancement. These techniques rely on the fact that the signal intensity and contrast of an image are largely determined by the data in the center of k-space (► Chapter 5.3) (► Fig. 45). All of the central k-space data can be acquired at the beginning of the scan by initially applying only shallow phase- and slice-encoding gradients. Sampling of the center of k-space while all of the contrast medium is still in the arterial system allows generation of images with good contrast and minimal venous overlap. This also applies when the peripheral k-space lines are filled after a considerable amount of the contrast medium has reached the veins. With this kind of centric k-space ordering, longer scan times and improved image quality are possible without compromising arterial contrast. Commercially available implementations of this technique are known as CENTRA or elliptical centric ordering of k-space.

It is crucial to collect the central k-space lines when the contrast medium concentration in the target vessels is highest. Several strategies are available for *optimizing bolus timing*:

— The test bolus technique is a method in which the individual patient's circulation time is determined by measuring the time the contrast medium needs to pass from the site of injection to the target vessel. To this end, a small amount of contrast (1 to 2 ml followed by a saline flush) is injected and the target area is repeatedly imaged using a fast sequence, e.g. a spoiled T1-weighted 2D GRE sequence which updates images once every second. The bolus must be large enough to cause signal enhancement when it arrives in the target vessel but should not unduly enhance the background signal in the subsequent 3D data acquisition. Based on knowledge of the individual circulation time determined in this way and the method of k-space ordering used, the 3D angiography sequence can be optimally coordinated with the injection of the contrast medium.

— Automatic triggering techniques are based on the continuous measurement of the vascular signal in a proximal test volume. Tracking starts with the injection of the angiographic bolus of the contrast medium, and the 3D sequence is then automatically triggered with an operator-controlled delay as soon as the signal intensity in the region of interest increases above a defined threshold. When the renal arteries are imaged, the test volume can be placed in the abdominal aorta.

— In a similar manner, 3D acquisition can be started manually as soon as the operator observes the arrival of contrast in the target volume on rapidly updated 2D images. This method is occasionally referred to as *fluoroscopic triggering*.

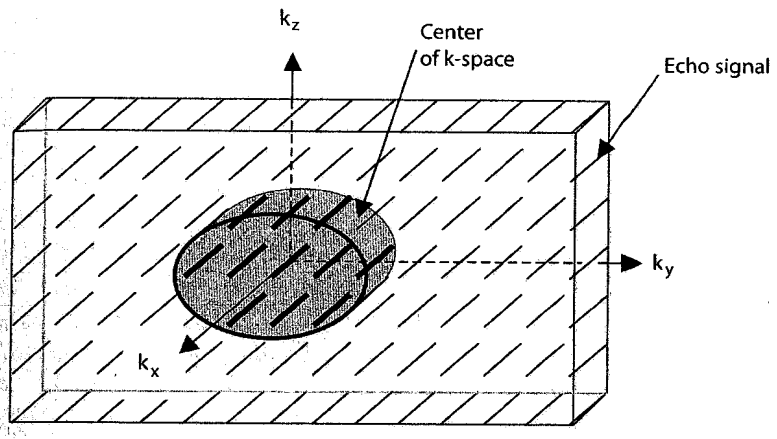

Fig. 45. Schematic representation of the MR raw data in k-space for a three-dimensional image. Each diagonal line represents an echo signal that is recorded within 1 or 2 msec. The contrast of the resulting MR image is mainly determined by the data in the center of k-space

Automatic or manual triggering techniques provide images with optimal arterial contrast when combined with a k-space ordering technique that samples the central lines first. However, these techniques are prone to early or late mistriggering of data acquisition. Moreover, rapid instruction of the patient is needed if breath-hold imaging is necessary. Bolus timing, on the other hand, is compatible with any method of k-space filling.

Even better suppression of the background signal is often achieved when the angiographic data set is acquired twice with the same parameters before and after contrast medium injection. The unenhanced image, the so-called *mask*, is then subtracted from the contrast-enhanced image. The resultant difference images highlight the signal changes occurring after contrast medium administration.

Many studies have shown contrast-enhanced MRA to have a high diagnostic accuracy in comparison with conventional radiographic techniques or other reference modalities. Most of the problems that arise in routine clinical application are associated with proper timing of data acquisition relative to contrast medium injection. The problems may be merely technical in nature or due to individual variations in circulation times and contrast medium distribution. An aneurysm, false lumen, or arteriovenous malformation may not be completely filled with contrast medium at the time of scanning even when there is adequate enhancement of the rest of the arterial system. When multi-station bolus chase is used, confounding signals from bright veins on projections of the lower leg images may limit the evaluation of the arterial tree. This is a problem more likely to occur in patients with diabetes mellitus. Retrograde inflow of the contrast medium may impair the diagnosis of vascular occlusion. However, as with other techniques, contrast-enhanced MRA generally overestimates rather than underestimates stenoses. With time-resolved imaging (► Chapter 11.1.3), many of the problems associated with achieving optimal bolus timing can be overcome. Contrast medium injection is minimally invasive. MR contrast media are associated with a very low rate of adverse events and allergic reactions are rare (► Chapter 12).

The advantages of contrast-enhanced MR angiography include:
- short scan time,
- three-dimensional display of large volumes in any orientation,
- high SNR and good vessel contrast,
- no exposure to ionizing radiation,
- well-tolerated contrast medium,
- minimal invasiveness of contrast medium injection, and
- reasonable robustness of the method under routine clinical conditions.

11.1.2 Black Blood Imaging

Black blood MR angiography is an MRA technique in which the signal from flowing blood is suppressed rather than enhanced as it is in most conventional MRA techniques such as TOF MRA. The black blood effect results from the fact that the blood in the scan plane is replaced with fresh blood during scanning.

Black blood MRA sequences are well suited to evaluate the vessel walls and the myocardium. They provide information on wall thickness, the

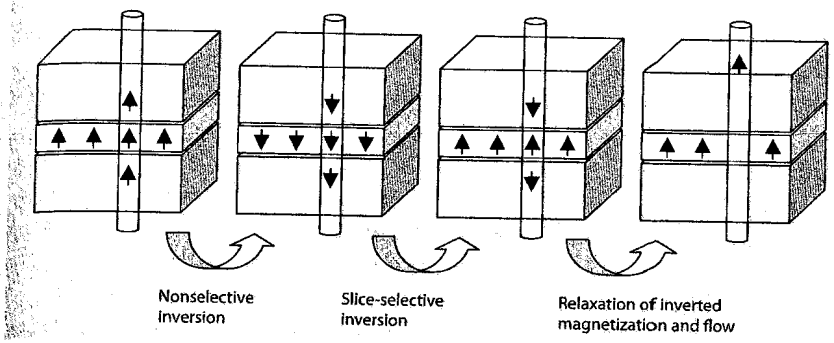

Fig. 46. Diagram of black blood MRA with double inversion recovery. The black arrows represent the longitudinal magnetization in the corresponding voxels

presence of inflammatory wall lesions, and the internal makeup of mural thrombi. So far, black blood MRA techniques have been mainly used to image large vessels such as the thoracic and abdominal aorta and the heart chambers or easily accessible vessels such as the carotid arteries. However, black blood MRA nicely depicts the coronary vessels as well.

The different effects of fresh blood flowing into the scan plane are mainly attributable to the fact that TOF imaging is performed with GRE sequences while SE sequences are used for black blood angiography. Blood whose magnetization is rotated into the transverse plane by the 90° excitation pulse of an SE sequence and then leaves the slice before the 180° refocusing pulse is delivered does not emit a signal (► Chapter 7.2). The two pulses are separated by half the echo time. Likewise, there is no signal from blood which is still outside the slice when the 90° RF pulse is applied but which then flows into the slice between excitation and readout.

The blood signal can be suppressed even more effectively by double inversion of longitudinal magnetization some hundreds of milliseconds before data sampling (double inversion recovery, ► Fig. 46). In this method, a non-selective 180° pulse, followed by a slice-selective 180° pulse, is applied to selectively rotate only the magnetization outside the scan plane into the negative z-direction. Magnetization relaxes and passes through zero before it regrows in the positive z-direction. Three conditions must be met for an improved suppression of the blood signal by double inversion recovery:
- The blood must be outside the scan plane during the two inverting pulses for its magnetization to be inverted.

— The blood must flow into the scan plane between double inversion and signal collection.
— Central k-space must be collected when the relaxing blood magnetization passes through zero. The interval between double inversion and the start of data collection is automatically calculated by the scanner's software.

Double inversion recovery can be combined with an additional inversion pulse to selectively rotate the longitudinal magnetization of fat into the negative z-direction prior to scanning. This will additionally suppress the signal from fat, as with a STIR sequence (► Chapter 7.5).

In the routine clinical setting, only 2D implementations of black blood MR angiography are available. The signal from slowly flowing blood as in the trabecular structures near the walls of the cardiac chambers may be difficult to suppress. The use of SE sequences makes the method somewhat slower than GRE-based techniques. Black blood MRA is an angiographic technique in the true sense of the word in that it primarily visualizes the vessel walls rather than the blood. The diagnostic accuracy of black blood angiography is not impaired by turbulent flow and the method has a lower rate of false-negative results in the evaluation of atherosclerotic lesions, especially in patients with early disease before significant narrowing of the vessel lumen has occurred.

11.1.3 Time-Resolved MR Angiography

The term *time-resolved MR angiography* is now mostly used to refer to the dynamic study of the distribution of a contrast agent in the vascular system. Technically, this is done by imaging a vascular region rapidly and repeatedly after administration of a single dose of contrast medium. The individual MRA images obtained in this way represent different phases of the progressive contrast medium distribution.

Ideally, time-resolved MR angiography depicts the early phases of contrast medium inflow, when all of the contrast medium is still confined to the arteries, and the subsequent venous phases when there is contrast medium in both the arteries and the veins. Time-resolved angiography can also encompass evaluation of organ perfusion, as has been shown for the kidneys.

When time-resolved MRA images are updated fast enough, arteries and veins are easier to distinguish even in case of suboptimal timing of data acquisition. Moreover, the method demonstrates the false lumen in dissection and facilitates the identification of retrograde contrast medium inflow.

Finally, the time-resolved information enables detailed evaluation of the vascular system supplying and draining an arteriovenous malformation or a tumor.

The demands on scan time are even higher for time-resolved MRA compared with contrast-enhanced MRA. To minimize scan time is a major concern but is usually achieved only at the cost of spatial resolution. The strategies available to shorten scan time are specific for dynamic imaging. A widely used approach is the reconstruction of data sets for which the periphery of k-space has not been updated (TRICKS or time-resolved imaging of contrast kinetics, keyhole imaging). In this method, peripheral k-space data from an earlier measurement is combined with the central k-space data which is updated more frequently. In the resulting images, the data from the center of k-space reflects the most recent change in signal intensities. The three-dimensional k-space used in this technique is divided into different areas where the image information is updated at different intervals. Data is updated more frequently, the closer the area is to the center of k-space. When this technique of k-space filling is combined with the methods of reducing scan time discussed earlier, 3D data sets can be acquired in 1 to 6 sec, depending on the size of the imaging volume and the desired resolution.

If even faster image acquisition is desired, one can dispense with phase encoding in the slice-select direction. In this way, one obtains two-dimensional images that represent projections of the signal intensities through the scan volume, similar to conventional X-ray techniques. Depending on the situation then, images can be updated several times per second with good spatial resolution.

11.2 Perfusion-Weighted Imaging

MR techniques that depict the flow of blood through the capillary circulation of an organ or tissue by different signal intensities are known as perfusion-weighted imaging (PWI). Perfusion-weighted images provide direct information on tissue perfusion, regardless of whether blood is supplied through the main vessel or collaterals. Perfusion imaging is mainly used to assess blood flow in the brain, the myocardium, the lungs, and the kidneys. Blood flow is measured in vivo by monitoring the signal changes that are induced by a tracer entering the tissue of interest. Exogenous and endogenous tracers are distinguished.

An example of exogenous tracers are the gadolinium-based contrast

agents used in contrast-enhanced MRA. These agents have very strong effects on the tissue signal when they flow into the target organ so that regional differences in perfusion are directly seen on the images (first-pass imaging).

The blood itself can be used as an endogenous tracer. To this end, the longitudinal magnetization of the blood in a feeding artery is saturated or inverted (arterial spin labeling, ASL). When the labeled blood arrives in the target anatomy before complete relaxation of its magnetization has occurred, it produces a decrease in signal. Because the signal decrease caused by the inflowing blood is usually too small to be seen directly, the contrast is highlighted by means of image subtraction using two sets of image data obtained with and without presaturation of the inflowing blood.

A paramagnetic contrast agent passing through a tissue induces transient shortening of its relaxation times, which is seen as an *increase in signal on T1-weighted images* and a *decrease on T2- or T2*-weighted images*. Both effects are exploited in MR imaging.

Contrast medium-based perfusion imaging of the heart, lungs, and kidneys is typically performed with T1-weighted GRE sequences. For cardiac perfusion imaging, the sequence must be synchronized with the cardiac cycle and generate at least one image from exactly the same phase of the cardiac cycle every second heartbeat. Sequences that collect more than one echo per excitation (multishot echo planar imaging, ▶ Chapter 8.5) are preferred due to their short scan time. Perfusion imaging of the lungs and kidneys is usually performed using T1-weighted 3D GRE sequences. Evaluation of contrast medium arrival in the target anatomy can be supplemented by monitoring the rate of contrast outflow from the tissue. Naturally, this additional step results in longer scan times.

Cerebral perfusion imaging is more commonly done with T2*-weighted 2D or 3D echo planar sequences which depict the passage of the contrast medium as a transient decrease in signal intensity (dynamic susceptibility contrast-enhanced MR imaging). With these sequences, most of the brain can be imaged with acquisition of a new image about once every second.

Ideally, one would determine absolute blood flow per unit time for each voxel of the target anatomy, for example, in milliliters per second and gram of tissue. In this way, one could identify even small areas with reduced flow relative to their surroundings and thus reliably diagnose globally reduced perfusion of an organ. Unfortunately, absolute quantification of blood flow is difficult to accomplish with both exogenous and endogenous tracers although many published studies report absolute values. Numerous factors have to be taken into account with both techniques and a review of the most

recent literature suggests that there is still no agreement as to the most suitable approach, at least with regard to the contrast medium-based methods.

Given these problems with absolute quantification of blood flow, various parameters have been proposed to characterize signal changes descriptively. Several of these parameters have been shown to be reproducible when repeat measurement is performed. Perfusion parameters determined by dynamic contrast-enhanced MRI include the time to peak signal enhancement, measured from the moment the first change is observed, or the signal change over time (enhancement slope). Although such parameters allow quantitative data analysis and are largely examiner-independent, they are nevertheless limited because results vary with the pulse sequence used and with other scan parameters as well. Hence, these parameters have to be calibrated after each change in the experimental setup and results are difficult to compare among different study centers.

Compared with various other modalities, MR perfusion techniques have the advantage of allowing noninvasive or minimally invasive evaluation of blood flow in a tissue with good spatial resolution. MR imaging involves no radiation exposure and is relatively fast. Patients can therefore be examined repeatedly, for example, to monitor therapy or to follow up surgery. Moreover, perfusion measurement can be performed in conjunction with other MR measurements in a single session. The additional morphologic data may provide detailed anatomic information or help differentiate vital regions of decreased perfusion from scar tissue or areas of acute infarction.

11.3 Diffusion-Weighted Imaging

Diffusion-weighted imaging (DWI) shows the changes in signal intensity resulting from the motion of water molecules by diffusion. Specifically, the signal of a biological tissue or body fluid is determined by the mean distance a hydrogen molecule moves per unit time based on random microscopic translational motion. The signal loss produced by the translational molecular movement in an MR image increases with the speed at which the molecules move through a magnetic gradient field. The direction and amount of diffusion weighting can be controlled by the operator by varying the direction and strength of the gradient field applied.

The motion of the water molecules is described quantitatively by the diffusion constant and usually varies with the direction of diffusion.

Isotropic diffusion is present when the distance traveled by the water molecules is the same in all directions (▶ Fig. 47). In an isotropic medium,

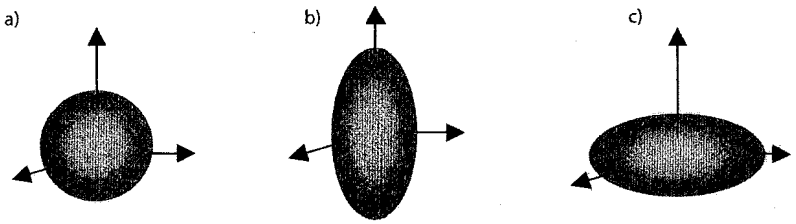

Fig. 47a–c. Diffusion tensor ellipsoids for isotropic (**a**), tubular (**b**), and layered environments (**c**)

the effects of molecular motion on the resulting MR images are independent of the direction of the gradient field. In the human body, nearly isotropic diffusion occurs in body fluids with freely mobile water molecules such as cerebrospinal fluid (CSF) in the ventricles or cystic fluids. The diffusion constants of such tissues are rather high and are identical in all directions. This results in a strong signal attenuation on diffusion-weighted images.

In an environment that is heterogeneous at the microscopic level, diffusion of water molecules is a directionally dependent phenomenon known as anisotropy. In the brain, for example, water molecules diffuse faster in the direction of axons with intact myelin sheaths than perpendicular to the axons. The diffusion constant is higher along the longitudinal axis of the axons than in the plane perpendicular to the axis. The diffusion-induced signal loss is smaller when the diffusion gradient is applied in a direction perpendicular to the fiber tract and larger when it is applied along the axis.

Diffusion of water molecules in an anisotropic medium is constricted by structures that are below the resolution of an MR image. Therefore, the directional differences can be observed only when most axons in a voxel are arranged in parallel and their effects add together.

Diffusion-weighted images depict lesions caused by stroke already within the first 6 hours of the onset of symptoms – before traditional MRI techniques such as T2-weighted images will show any significant changes. In the acute phase, the diffusion-induced loss of signal is less pronounced in affected areas and these appear brighter compared with unaffected brain. This positive contrast is gradually lost in the course of some days and finally becomes negative as a result of greater mobility of the water molecules.

Diffusion-weighted images are typically acquired with an echo planar

imaging technique. A pair of gradient pulses is delivered between the excitation pulse and signal collection to sensitize the sequence to diffusion effects (▶ Fig. 48). The pulse pair differs from that used in phase-contrast angiography in that both halves have the same polarity. However, the effect is very similar due to the 180° RF pulse which is delivered between both halves of the pulse. A change in phase is imparted only to those spins that move along the gradient field while the pulses are being applied. As a result, the spins in a voxel which have experienced different phase shifts are no longer coherent and produce a weaker MR signal. The signal attenuation depends on the strength and duration of the gradient pulses, their spacing, and the diffusion constant along the direction of the gradient field.

The amount of diffusion weighting achieved with a given gradient pulse pair and inversion pulse sandwich is denoted by the b-value. This factor expresses the signal loss to be expected from a given pulse sequence for a given diffusion constant.

Diffusion constants in biological tissues can be measured by repeated scanning with different b-values but otherwise identical imaging parameters, in particular an unchanged gradient direction. The measured diffusion constants are represented by the *apparent diffusion coefficient* (ADC), which is distinct from the constant of unobstructed diffusion in pure water.

Images whose gray-scale values represent the mean ADCs of the corresponding voxels are known as *ADC maps*. An area of acute infarction that is bright on a diffusion-weighted image (reduced mobility of the water molecules) will appear dark on the corresponding ADC map (smaller diffusion constant).

Diffusion constants for different directions can be measured by changing the direction of the gradient field. Such measurements provide detailed information on the local geometry of the microscopic structures that restrict

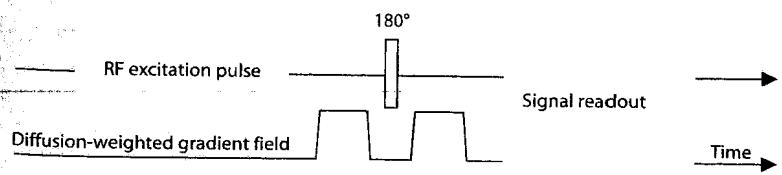

Fig. 48. Diagram of a diffusion-weighted sequence

water diffusion. Based on the measurement of the diffusion constants in six selected directions, the entire geometry can be calculated by using the formalism of three-dimensional tensors. This version of diffusion imaging is known as *diffusion tensor imaging* (DTI). Such a formalism provides an approximation of the mean diffusion of water molecules in all directions in an ellipsoid whose three main axes may differ in length when there is anisotropic diffusion (► Fig. 47). A more accurate geometric model of the structures that hinder diffusion in a voxel can be generated when additional diffusion constants for other directions are measured.

Diffusion tensor imaging is mainly used for so-called fiber tracking (tractography) in the cerebral white matter. The information obtained with DTI is used to reconstruct the spatial course of fiber tracts over longer distances from the relative orientation and size of diffusion ellipsoids in adjacent voxels.

Diffusion-weighted images are highly sensitive to all kinds of movements. These include rotation or trembling of the head in cerebral imaging or respiratory motion in imaging of the trunk. This is why short scan times are important. Fast switching of the strong gradient pulses requires a powerful MR scanner. When a sequence is made sensitive only to diffusion in a specific direction, normal areas may show false positive contrast if the dominant orientation of the fiber tracts is perpendicular to the preselected diffusion direction. The radiologist interpreting the images should therefore take into account information on diffusion in 3 orthogonal directions, which can be obtained with a single scan.

The gradient pair applied to make the sequence sensitive to diffusion processes only attenuates the signal compared to images obtained without the gradient. Structures such as CSF with a strong signal on corresponding non-diffusion-weighted images may still appear bright on images with only mild to moderate diffusion weighting when their diffusion constant is high. This effect is known as *T2 shine-through* and may be difficult to distinguish from actual restriction of diffusion. Only on strongly diffusion-weighted images are the signal intensities predominantly determined by diffusion.

Diffusion-weighted imaging is an area of intensive research because it provides unique information that cannot be obtained with other methods or only to a very limited extent.

11.4 The BOLD Effect in Functional Cerebral Imaging

Functional magnetic resonance imaging (fMRI) of the brain aims at identifying cerebral areas that respond to a well-defined external stimulus by a change in signal (brain mapping). Functional images are typically acquired using T2*-weighted techniques. Classical tasks used to induce neuronal responses are visual (such as looking at changing patterns) or sensorimotor (such as a sequence of defined finger movements) activation. A wide variety of protocols exist for neuronal activation and the interpretation of the changes observed on functional MR images (paradigms).

Functional MRI is based on the assumption that a stimulus increases the oxygen demand of a specific brain region that is activated by it. To meet the higher demand, capillary blood flow and the blood volume in the activated region are increased by local vasodilatation. Moreover, it is assumed that excess oxygen is supplied to the activated area because the increased blood flow exceeds the metabolic needs after some time. The higher proportion of hemoglobin molecules bound with oxygen (oxyhemoglobin) prolongs the T2* time of the surrounding water, which is observed as a signal increase on T2*-weighted images. This contrast mechanism is known as blood oxygen level-dependent (BOLD) contrast.

The T2* relaxation rate of blood depends on whether or not the hemoglobin is bound with oxygen. Hemoglobin not combined with oxygen (deoxyhemoglobin) is paramagnetic because of unpaired electrons and shortens the T2* of surrounding water. In contrast, oxyhemoglobin is slightly diamagnetic because all electrons are paired and thus has only a negligible effect on the relaxation times of surrounding water. This is how an increased oxygen saturation lengthens the T2* of blood water.

While BOLD imaging is based on the oxygen content of blood, there are other functional MRI techniques that take advantage of the higher blood flow or the increased blood volume to demonstrate cerebral activation.

BOLD imaging is typically performed with strongly T2*-weighted GRE EPI sequences (► Chapter 8.5) that allow scanning of the entire brain in a few seconds. To capture the fairly small signal changes induced by activation, all slices are usually imaged repeatedly. Imaging is continued for some time with alternating "on" and "off" cycles (block design paradigm, ► Fig. 49). Interpretation of the data requires sophisticated statistical methods to correlate the signal changes on the MR images with the task paradigm presented. In this way, maps of brain activation are generated where

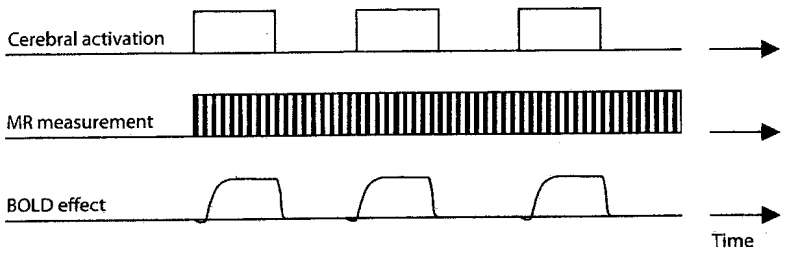

Fig. 49. Block design paradigm for functional brain imaging

voxels, identified as representing real activation by the application of statistical thresholds, are colored. The final activation maps are superimposed on traditional morphologic MR images that depict anatomic structures with a higher resolution and thus allow exact identification of the brain areas being activated.

The BOLD contrast increases with the magnetic field strength of the MR scanner. The noise that is associated with MR scanning makes it somewhat difficult to measure cerebral activation by auditory stimuli. Moreover, the standard techniques of stimulation have only a limited temporal resolution for the registration of physiologic changes. Therefore, event-related paradigms with only short periods of activation are becoming more popular. The spatial resolution of BOLD imaging is limited because the area with an increased oxygen saturation of the blood may be much larger than the region that has actually been activated. Finally, T2* is affected by many confounding factors at the microscopic level that may be difficult to isolate. This is why the magnitude of the observed signal change does not provide a quantitative measure of the physiologic changes induced by stimulation.

While much research activity is focused on functional MR imaging, it has only a very small role in routine clinical applications at most radiologic centers. Clinically, BOLD imaging is used to plan neurosurgical interventions. Despite its limitations, functional BOLD imaging enables fully noninvasive and radiation-free evaluation of subtle changes in cerebral activity with a spatial resolution of 1–2 mm or better and a temporal resolution in the range of 100 msec.

11.5 Cardiac Imaging

Imaging of the heart differs from imaging of other organs in that the constant cardiac motion causes blurring and other artifacts along the phase-encoding direction on MR images acquired with long scan times. With state-of-the-art equipment, though, the scan time for acquisition of a single slice can be reduced to such an extent that cardiac motion can be monitored on a series of images in near-real time without degradation of image quality by artifacts caused by respiratory or cardiac motion. Most artifacts can be effectively eliminated when the scan time is less than 50 msec during systole and less than 200 msec during diastole.

Real-time cardiac imaging is mainly performed to rapidly localize the heart and long- and short-axis views for subsequent data acquisition.

To improve the spatial or temporal resolution of "real-time" cardiac imaging, an image is obtained over several heartbeats (so-called segmented acquisition). This is possible because cardiac motion is periodic under normal conditions and the myocardium will be in the same place at specific time points within different cycles. To ensure that all data for an image is sampled during the same phase of the cardiac cycle, segmented imaging must be tailored to the individual patient's heart rate. To this end, an electrocardiogram (ECG) is recorded and the data is used by the scanner software to identify the R wave in each cardiac cycle. The ECG data can be used in two ways, either to trigger MR acquisition to a specific phase of the cardiac cycle (cardiac triggering, prospective cardiac gating) or to retrospectively assign continuously acquired data to the corresponding cardiac phases (retrospective cardiac gating).

The scan time per cardiac cycle is shorter when an image is acquired over several cycles, resulting in an improved temporal resolution of cardiac motion. However, the overall scan time per image is longer and effects resulting from respiratory motion between scans become stronger.

These limitations can be overcome by breath-held imaging. For instance, with a repetition time (TR) of 3.5 msec and collection of only one phase-encoding step per R-R interval, 14 phase-encoding steps can be sampled in 50 msec. If we want to generate an image with a resolution of 224 pixels in the phase-encoding direction, the acquisition will have to be segmented to

224/14 = 16 heart beats. However, patients with heart disease may find it difficult to hold their breath for 16 heart beats.

Since respiratory motion is also periodic, data acquisition cannot only be distributed over several cardiac cycles but also over several breaths. This is accomplished by monitoring the patient's breathing rhythm: A short 1D scan is alternated with image data acquisition for localizing the boundary between the diaphragm and the lung along the body's longitudinal axis. In this way, the image data can be – prospectively or retrospectively – assigned to the different phases of the respiratory cycle (navigator technique). Using the navigator technique, scanning is not limited to the duration of a breath-hold but can be performed with the patient breathing freely. Navigator techniques are limited by rather inefficient acquisition and long scan times. Moreover, they provide the best results in healthy subjects with a fairly regular heart rate and breathing pattern.

When used in combination with gating, the cardiovascular MR imaging techniques described so far allow three-dimensional visualization of the anatomy of all cardiac chambers and of the vessels entering and leaving the heart, without radiation exposure and with generally good sensitivities. MRI can thus be used to repeatedly examine patients with suspected congenital malformations, cardiomyopathy, valve incompetence, or pericardial disorders; to follow up patients after bypass surgery; and to monitor heart transplant recipients. A wide variety of pulse sequences and sequence modifications are in use for imaging of the coronary vessels, all having specific advantages and disadvantages. The major advantages of cardiac MR imaging lie in the repeat evaluation of morphology, function, and perfusion without radiation exposure in patients with coronary heart disease as well as in the localization and precise delineation of infarcted areas.

Some specific applications are discussed in more detail below.

11.6 Cardiac Imaging with SSFP Sequences

Steady-state free precession imaging has become a fixed component of standard cardiac MRI protocols. With its shorter TR (about 2-5 msec) compared with other GRE sequences, an SSFP sequence (► Chapter 7.7) yields images with a stronger blood signal. It is thus possible to rapidly image the blood in the cardiac chambers with good contrast relative to the myocardium. Good contrast is even achieved when there is only little blood flow in the scan

plane and the blood signal is not enhanced through inflow effects. This may be advantageous when longitudinal views of the left ventricle are obtained.

The sequence is usually acquired and displayed in the cine mode with imaging of each slice during different phases of the cardiac cycle. If, for instance, we assume an acquisition with scan time segments of 50 msec and a patient with a heart rate of 70 beats per minute, cardiac motion could be assessed on a sequence of 17 images from different phases of the cardiac cycle obtained with a single acquisition. When several slices are acquired in this way during different breath-hold periods, the motion of the entire heart can be evaluated and quantified. Even the apex of the heart is depicted with good quality on long axis views.

The acquired image data sets can be used to determine global morphologic and functional parameters such as the myocardial mass, the ejection fractions of both ventricles, or the stroke volume. These parameters can be determined directly without having to make geometric assumptions as in the classical model-based methods. There is good interobserver reproducibility of the results.

Besides estimation of the global parameters, the method provides information on regional functional parameters such as local wall motion or left ventricular wall thickening from diastole to systole. Disturbed perfusion can be diagnosed with a high degree of accuracy if a myocardial region showing normal wall motion at rest becomes hypokinetic during drug-induced stress (dobutamine).

A slight additional enhancement of the blood signal on SSFP images can be achieved by administration of moderate amounts of an MRA contrast agent.

SSFP images are degraded by inhomogeneities in the static magnetic field, especially in connection with flow effects, and an inadequate RF frequency. Nevertheless, the technical problems have been solved to such an extent that SSFP has become very reliable for routine clinical application.

11.7 Myocardial Perfusion Imaging

Myocardial perfusion is typically evaluated as the signal enhancement seen on T1-weighted MR images obtained during the first pass of a contrast medium through the muscle tissue. Ideally, the image is updated once every heart beat. The contrast medium is injected intravenously, usually at a lower

dose than administered for angiography. Ischemic areas are identified directly by a delayed inflow of contrast medium and/or a lower peak signal intensity during passage of the contrast medium. The differences to adjacent myocardium with normal perfusion are especially salient when viewing the images in rapid succession in the cine mode. In this way, it is also possible to identify disturbed perfusion confined to inner myocardial layers, which is more difficult to detect with competing diagnostic modalities.

Imaging is performed during drug-induced stress (adenosine, dipyridamol) and breath-hold. The actual scan takes less than a minute. Regions of reduced perfusion are differentiated into viable and nonviable areas by combining stress imaging with perfusion measurement at rest or with late-enhancement imaging (▶ Chapter 11.8).

The most widely used techniques for myocardial perfusion imaging are fast GRE and multishot EPI in conjunction with a preparatory RF pulse. The preparatory pulse is either a 90° saturation pulse or a 180° inversion pulse, resulting respectively in a saturation recovery sequence or an inversion recovery sequence. The latter allows stronger T1 weighting while the former is more stable in that it is less sensitive to an irregular heart rate and yields more reproducible results. With optimal parameter settings and the options available on specific scanners, it is currently possible to acquire about four slices per heart beat or eight slices every second heartbeat.

For quantitative analysis, the temporal course of the contrast medium concentration in the myocardium is related to the temporal course of the concentration in the blood in the feeding arteries. Since the course cannot be measured directly for each voxel, an approximation is used to determine the course in the blood in the left ventricle. This technique of quantitative analysis is associated with a number of problems: different signal enhancement in myocardium and blood, unclear effect of water exchange through cellular and capillary walls, unknown patency of the capillary membrane for the contrast medium, signal differences due to local variations in sensitivity of the receive coil, and nonquantifiable signal enhancement resulting from respiratory motion of anatomic structures in the imaging plane. Various options are available to tackle these problems.

Despite these problems, results reported in the literature suggest that quantitative data obtained with this technique is fairly independent of the examiner and has a high diagnostic accuracy compared with different reference modalities.

11.8 Late-Enhancement Imaging

On late-enhancement images acquired about 10 to 20 minutes after intravenous administration of an angiographic contrast agent dose, a bright signal indicates a myocardial area of increased contrast accumulation relative to surrounding normal myocardium. In this way both acutely infarcted tissue and scar tissue after an older infarction can be delineated with good resolution ("bright is dead"). Increased contrast accumulation in these areas is attributed to a larger extravascular, extracellular volume and/or slower washout. Studies performed so far suggest that late-enhancement images allow very accurate estimation of the size of an infarcted area.

Late enhancement is not a specific feature of myocardial infarction. A similar signal enhancement may also be seen in myocardial regions affected by other heart diseases. While enhancement associated with infarction is usually confined to subendocardial regions with transmural extent in severe cases, the late enhancement seen in other disorders may be confined to the middle layer of the wall.

It must be noted, however, that very poorly perfused, nonviable areas may not show contrast enhancement due to failure of the contrast medium to enter these areas by the time the images are acquired. This applies especially to images that are obtained within the first minutes of contrast medium administration. Enhancement may be absent in very large infarctions where the center appears dark while the periphery is bright. In case reports in the literature, this phenomenon is described by such terms as "microvascular obstruction". It has been shown that unchanged microvascular obstruction persisting for several days is associated with an extremely poor prognosis.

Late-enhancement images are acquired with GRE-based inversion recovery sequences. The recovery time between the RF inversion pulse and data acquisition (*TI*, inversion time) is selected such that the magnetization of healthy myocardium passes through zero when the central k-space lines are filled, leaving normal tissue dark on the resultant image. If a scan takes several minutes, it may become necessary to readjust TI to the changing contrast medium concentration. Late-enhancement imaging can be performed with 2D or 3D sequences.

While the differentiation of infarcted and healthy tissue has always been straightforward, the differentiation of a subendocardial infarction and blood in the left ventricle may be difficult and require additional scans, for example with a different *TI*.

It has been suggested that late-enhancement imaging has the potential to become the method of first choice for demonstrating myocardial infarction and estimating its extent.

11.9 Detection of Increased Myocardial Iron Concentrations

MR imaging seems to have the potential to reliably detect excessively high iron concentrations in the myocardium on the basis of their T2*-shortening effect when precisely defined protocols are used for data acquisition and analysis. Such protocols comprise a short axis slice through a central portion of the left ventricle which is obtained repeatedly with different echo times using a GRE sequence. The signal loss observed with increasing echo times allows one to calculate the T2* relaxation constant in a region of interest placed in the septum. Preliminary results, mostly obtained in thalassemia patients, suggest that T2* shortening predicts a deterioration of cardiac function only if the value drops below a certain threshold. The threshold identified is about 20 msec at 1.5 tesla compared with a mean value of 52 msec in healthy subjects. Measurement of T2* relaxation times might therefore provide the basis for identifying those patients who will benefit from intensive iron-chelating therapy and so be spared the poor prognosis associated with impairment of cardiac function.

12 MR Contrast Agents

JOHANNES M. FROEHLICH

Image contrast in medical MR imaging results from differences in signal intensity (SI) between two tissues and is determined by intrinsic and extrinsic factors. These are respectively properties of the different tissues and properties of the MR scanner, especially of the pulse sequence used.

MR contrast media are pharmaceutical preparations that are administered in MR imaging to further enhance the natural contrast and additionally to obtain dynamic (pharmacokinetic) information. To achieve these goals, contrast agents used for MRI must have specific physicochemical properties and also a suitable pharmacokinetic profile.

MR contrast media fundamentally alter the intrinsic contrast properties of biological tissues in two ways:

- *directly* by changing the proton density of a tissue or
- *indirectly* by changing the local magnetic field or the resonance properties of a tissue and hence its T1 and/or T2 values.

The local magnetic field strength is altered because the unpaired electron spins of the contrast medium (CM) interact with the surrounding hydrogen nuclei of the water, fat, or protein molecules in the tissue. Thus, the mechanism of action of an MR contrast agent comprises processes of the electron shell and not just processes at the nuclear level, as does the MR effect. The magnetic moments of electrons are 657 times greater than those of protons. This is one of the reasons why the electron shell has much more powerful paramagnetic properties than a hydrogen nucleus.

The interactions occurring between contrast medium electrons and tissue protons comprise "inner-sphere relaxation" (through interaction with bound water) and "outer-sphere relaxation" (e.g. arising from the diffusion of water nearby). Both processes contribute substantially to the overall effect of MR contrast media.