Magnetic Resonance Imaging (MRI) I

Physics	Modern Physics	Quantum physics	
Biology	Modern Imaging Methods in Biology		
Applied Science	Medicine	The Nervous System	
Applied Science	Medicine	Radiology & Ultrasonic Diagnostics	Magnetic resonance imaging (MRT)

Difficulty level	Group size	Preparation time	Execution time
hard	2	10 minutes	20 minutes

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General information

Application

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Setup

Nuclear magnetic resonance (NMR) is a physical phenomenon in which nuclei in a strong constant magnetic field are perturbed by a weak oscillating magnetic field (in the near field) and respond by producing an electromagnetic signal with a frequency characteristic of the magnetic field at the nucleus. It has many applications in fields such as medicine where it is just as a tool for imaging.

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Other information (1/4)

Prior knowledge

The prior knowlege required for this experiment is found in the theory section.

Main principle

The aim of these experiments is to study the fundamental principles of 2D magnetic resonance imaging by way of two methods that are based on different gradient techniques. The experiments are performed directly with the MRT training unit. This unit enables the direct examination of small samples in a sample chamber. The unit is controlled via the supplied software. The examinations include the generation of a two-dimensional sectional image of a sample by way of the spin echo method, which involves the examination of a spin echo signal that is superimposed by a magnetic gradient field, and the generation of a two-dimensional sectional image of a sample by way of the gradient echo method in which a suitable gradient echo signal is examined. The gradient echo method is closely joined to the extremely fast FLASH method (Fast Low Angle Shot).

Other information (2/4)

Tasks

A: Application of frequency and phase encoding for the generation of a spin echo signal based on which a 2D image can be acquired (Spin Echo 2D)

1. Record a transverse (X-Z) 2D MR image of a double sample of oil and water by way of the spin echo method. In a transverse 2D MR image, the cross-sectional plane is parallel to the bottom of the magnetic resonance tomography scanner.
2. Generate 2D MR images of the double sample with the settings of 1 for all of the three slices that can be selected (X-Y, X-Z, Y-Z). Please note that the generated images are accumulated images over the entire sample space, which means that the selected slice always has the thickness of the entire sample space.
3. Study the effects of the number of data points and phase steps on the transverse 2D MR image of the double sample.

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Other information (3/4)

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Tasks

4. Record various transverse 2D MR images of the double sample with different strengths of the readout and phase gradients.
5. Record various transverse 2D MR images of the double sample with different repetition and echo times.
6. Study the influence of the number of averages on the transverse 2D MR image of the double sample of oil and water.
7. Record a transverse (X-Z), sagittal (Y-Z), and coronal (X-Y) 2D MR image of the structure sample. Use the parameter settings of 1.

Other information (4/4)

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Tasks

B: Application of frequency and phase encoding for the generation of a gradient echo signal based on which a 2D image can be acquired (Flash 2D)

1. Record a transverse 2D MR image of a double sample of oil and water by way of the rapid gradient echo method. In a transverse 2D MR image, the cross-sectional plane is parallel to the bottom of the magnetic resonance tomography scanner.
2. Record various transverse 2D MR images of the double sample with different repetition and echo times.
3. Study the influence of the deflection angle on the transverse 2D MR image of the double sample of oil and water.
4. Record a transverse (X-Z), sagittal (Y-Z), and coronal (X-Y) 2D MR image of the structure sample with acceptable quality. Try to produce these images as quickly as possible.

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Safety instructions

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- Read the supplied operating instructions thoroughly and completely prior to starting the unit. Ensure that all of the safety instructions that are listed in the operating instructions are strictly followed when starting the unit.
- Only use the unit for its intended purpose.
- Pregnant women as well as people with cardiac pacemakers must keep a distance of at least 1 m from the magnet.

Theory (1/31)

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In the experiment group "Fundamental principles of nuclear magnetic resonance (NMR)", we have seen, for the alignment of nuclear spins in a static magnetic field $\vec{B}_0$, that there is a slightly preferred alignment in parallel to the static magnetic field, which finally leads to a longitudinal magnetisation $\overrightarrow{M_{L0}(t)}$ parallel to $\vec{B}_0$ The nuclear spins then precess around the static magnetic field vector $\vec{B}_0$ with a frequency that is highly specific for the nucleus. This frequency is called the Larmor frequency. The following applies:

$V_L = \frac{\omega_L}{2\pi} = \frac{y}{2\pi} B_0$ (1)

An HF pulse with the frequency V_L, which is applied perpendicularly to $\vec{B}_0$, deflects the total magnetisation vector (resonance condition). In the case of a 90° excitation pulse, the initial longitudinal magnetisation $\overrightarrow{M_{L0}}$ is completely transformed into a transverse magnetisation $\overrightarrow{M_Q(0)}$. This transverse magnetisation then precesses around the static magnetic field vector with the Larmor frequency.

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Theory (2/31)

In the experiment group "Relaxation times in nuclear magnetic resonance", we were able to demonstrate that the relaxation of the deflected magnetisation vector back to its state of equilibrium is described by two relaxation times. While the exponential restoration of the longitudinal magnetisation $\overrightarrow{M_L(t)}$ is described by the relaxation time T_1, the exponential decay of the transverse magnetisation $\overrightarrow{M_Q(t)}$ is described by the relaxation time T_2. The following applies:

$M_L(t) = M_{L0}(1 - ce^{-t/T1})$ (2) or $M_Q(t) = M_Q(0)e^{-t/T2}$ (3)

with M_{L0} as the strength of the initial longitudinal magnetisation and $M_Q(0)$ as the strength of the transverse magnetisation directly after the HF pulse that was applied with the Larmor frequency. The formula $c = 1 - cos\varphi$ applies. Thus, in the case of a 90° excitation c equals 1.

Theory (3/31)

Based on various interactions, T_2 is normally smaller than T_1. The transverse magnetisation that decreases exponentially is the actual MR signal that can be detected by way of the receiver coils. This signal is called an FID signal (free induction decay). In the experiment group "Spatial encoding in nuclear magnetic resonance", we were able to see how the slices of the sample can be selected and how localised signals are assigned to these slices. These signals can be used to acquire a spatially resolved MR image by way of a Fourier transformation. The trick was to superimpose the static magnetic field $\overrightarrow{B_0}$ with magnetic gradient fields that are generated by opposite pairs of coils. The slice selection concept by way of a superimposed linear magnetic gradient field is clear. Since the static magnetic field is increased at one gradient coil and decreased at the opposite coil, the individual spins precess with different speeds at different locations. If the HF pulse is then applied precisely with the Larmor frequency ω_0, this pulse would excite the spins only at a certain resonance point z_0. This point is called the slice position (see Fig. 1). The thickness of the slice can be selected via a certain bandwidth of the HF pulse in combination with the strength of the gradient field (see Fig. 1). If no slice selection gradient is applied, the detected signal always corresponds to the sum of all of the signals over the entire relevant sample space.

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Theory (4/31)

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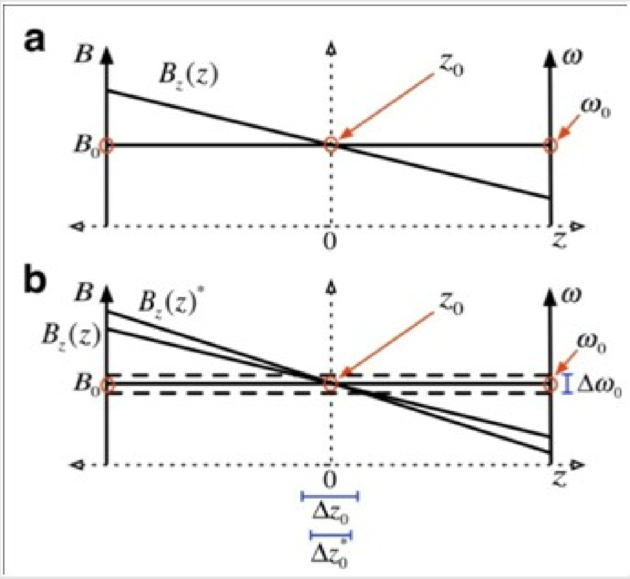

Fig. 1: Gradient field for the slice selection in the z-direction. Figure (a) shows that, due to the position-dependent gradient or due to the new, position-dependent magnetic field B_z (z), the nuclear spins come into resonance with the HF pulse with the frequency ω_0 only at the point z_0. Figure (b) shows how the thickness Δz_0 of a slice can be selected via the bandwidth $\Delta\omega_0$ of the exciting HF pulse and via the slope of the gradient.

Theory (5/31)

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Magnetic gradient fields also enable the assignment of signals to their exact location of origin, i.e. gradient fields enable the spatial encoding of a slice and, thereby, the actual generation of the MR image.
This is where the concept of pixels or voxels comes into play. Every MR image consists of individual image elements, the so-called pixels (2D) or voxels (3D). Since the selected slices always have a certain thickness in MR imaging, the term voxels is used in most cases. The individual voxels have characteristic grey values that are proportional to the signal strength. The resolution is the total number of voxels and, thereby, the total number of different grey values. The higher this number is the more pieces of signal information must be assigned to their location of origin. The process of this highly important signal information assignment shall be briefly described once again.
For this purpose, let us have a look at our spin echo signal that has already been presented in detail in the experiment groups "Fundamental principles of nuclear magnetic resonance" and "Relaxation times in nuclear magnetic resonance". For a start, we try to enforce 1D spatial encoding, i.e. the encoding of a voxel strip (see Fig. 2).

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Theory (6/31)

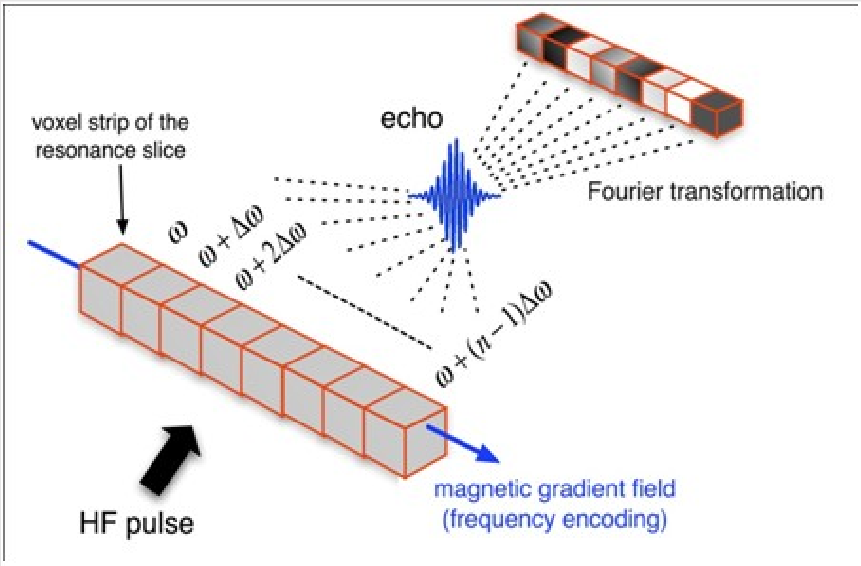

Fig. 2: Spatial encoding of a 1D voxel strip (n voxels). During the measurement of the spin echo, a linear magnetic field gradient is applied in the encoding direction. This gradient causes the nuclear spins of the voxel strip to precess with a frequency that increases linearly (frequency encoding). The echo signal is a mixture of all of the signals of each of the voxels (compare the sound of several different frequencies). Based on the signal mixture, the grey value of a voxel in the position space can be determined with the aid of a Fourier transformation.

Theory (7/31)

If we apply a linear magnetic field gradient in the encoding direction during the measurement of the spin echo, the nuclear spins in various voxels along the voxel strip precess with a frequency that increases linearly. The echo signal that can be measured is a mixture of all of the signals of each of the voxels of this voxel strip. A magnetic gradient thus applied is called a frequency encoding gradient and the corresponding method is referred to as frequency encoding (see Fig. 2). The signal strength of a certain voxel can be assigned precisely to this voxel based on the frequency and with the aid of a Fourier transformation. The result is a projection of the medium that is to be studied on the gradient axis. The signal strength determines the grey value of the associated voxel. This leads to a spatially resolved 1D MR image of a voxel strip (stripes with different grey values).
However, for the encoding of two-dimensional images, frequency encoding alone is not sufficient. This is due to the fact that during frequency encoding in both matrix directions, two voxels can have the same frequency in a 2D frequency matrix. This indiscernibility would render the MR image irreproducible. This is why spatial encoding in a second direction must be realised in a different way.

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Theory (8/31)

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If a linear magnetic field gradient is briefly applied between the HF pulse and the spin echo in the second direction of the image matrix that is to be analysed, the individual nuclear spins briefly precess at different speeds along the voxel strip in this direction. This is why the spins of the individual voxel elements have different phase positions (compare dephasing due to local magnetic field inhomogeneities) (see Fig. 3). In the case of a resolution of m voxel elements in the second encoding direction, m different linear magnetic field gradients must be applied in m consecutive measurement sequences (e.g. by a variation of the amplitude of the magnetic field gradient) that produce m spin echoes with different phase encodings. This is the only way to assign the recorded signals precisely to their location of origin also in the second encoding direction. The assignment is once again realised by way of a Fourier transformation that enables the phase positions of a signal to be filtered out. A magnetic gradient thus applied is called a phase encoding gradient and the corresponding method is referred to as phase encoding (see Fig. 3).
The clever application of magnetic field gradients and the subsequent recording of spin echo signals (their number is given by the resolution in the phase encoding direction) enable the reconstruction of a 2D MR image.

Theory (9/31)

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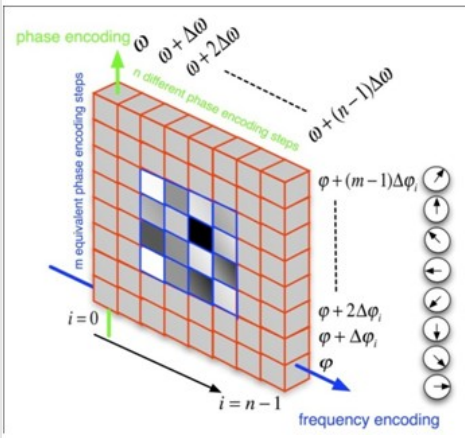

Fig. 3: Spatial encoding of a 2D voxel slice (n x m voxels). During the measurement of the spin echo, a linear magnetic field gradient is applied in one of the two encoding directions (e.g. in the x-direction). This gradient causes the nuclear spins along this direction to precess with a fre-quency that increases linearly (frequency en-coding, shown in blue). In the second encoding direction (e.g. in the y-direction), a magnetic field gradient is briefly applied prior to the measurement of the spin echo. This gradient applies different phase positions to the nuclear spins along the corresponding encoding direc-tion (phase encoding, shown in green). In order to guarantee the discernibility of the individual voxel signals, phase encoding must be repeated m times with different gradients. The grey values of the various voxels in the position space can be determined based on the mixture of signals of different frequencies and phases, and with the aid of a Fourier transformation.

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Theory (10/31)

Let us look at the example of a matrix of 64 x 64 voxels. In one direction, we obtain simple spatial encoding by way of a frequency encoding gradient. In the other direction, we need 64 spin echoes with different phase encoding, i.e. 64 phase encoding steps, for the spatial encoding of 64 voxels. The 64 echo signals are written line by line into a matrix, the so-called 2D raw data matrix of the spatially encoded time signals. The associated space is called k-space or wave-vector space. Each of the 64 x 64 points of the k-space corresponds to an angular wave number $\vec{k} = \omega / c$ with a well-defined direction. As a result, every point corresponds to a stripe pattern (see Fig. 4). Based on these spatial stripe patterns, an image can be composed. The raw data values of the k-space determine the weighting of the individual stripe patterns. Crude stripe patterns have a low spatial frequency close to the centre, whereas fine stripe patterns have a high spatial frequency and are located further outward in the k-space. Based on the raw data matrix, i.e. the weighting of the stripe patterns, a 2D Fourier transformation then calculates the grey value distribution in the position space. This means that the image is reconstructed by assigning a specific grey value to every voxel (see Fig. 4).

Theory (11/31)

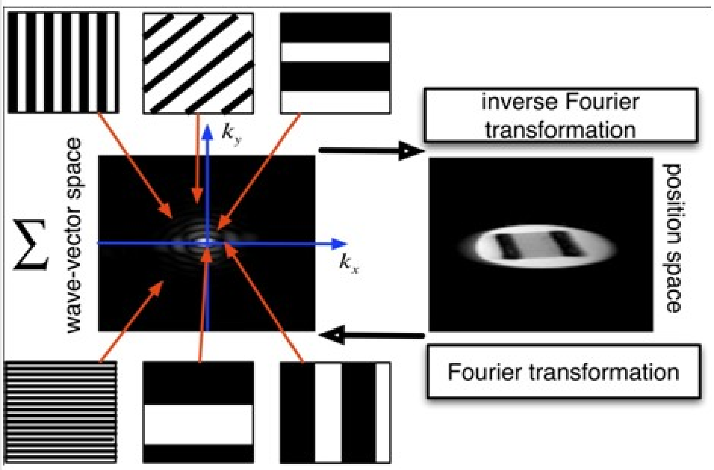

Fig. 4: 2D imaging in the k-space and position space. In the k-space, every point corre-sponds to a stripe pattern with a well-defined frequency. The raw data values in the k-space determine the weighting of these stripe patterns and, thereby, the wave-vector image. Coarser strip patterns are very close to the centre of the k-space. They determine the coarse structure and contrast of the image in the position space. Finer stripe patterns are located further outward in the k-space. These stripe patterns provide information concerning the borders, edges, contours and, thereby, of the resolution. The relationships between the position space and k-space are given by way of Fourier transformations.

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Theory (12/31)

We will now once again summarise the described procedure for magnetic resonance imaging based on the spin echo technique (Spin Echo 2D). In doing so, we will put special emphasis on the sequence that is used as well as on the most important parameters (see Fig. 5).
The starting point for our considerations is the state of equilibrium, i.e. the alignment of the magnetisation vector parallel to the static magnetic field $\overrightarrow{B_0}$. A 90° HF pulse that fulfils the resonance condition (1) deflects the magnetisation vector by 90° into the plane perpendicular to the static magnetic field $\overrightarrow{B_0}$. There, it continues to precess around $\overrightarrow{B_0}$until the nuclear spin ensemble dephases in the time $T_2^* < T_2$. The result is the FID signal. Since the transverse magnetisation does not relax until after the time T_2 and since the dephasing that evokes the FID signal is of a systematic nature, the lost signal can be restored within the time T_2 by way of a 180° pulse after the time \(T_S\) (compare spin echo). However, prior to the 180° pulse, a magnetic gradient of half the duration τ_{FG} of the final readout gradient ($2 \cdot \tau_{FG}$) (see below) is applied in the first encoding direction (e.g. in the x-direction).

Theory (13/31)

At the same time, another magnetic gradient with a certain amplitude that depends on the measurement cycle is applied in the second encoding direction (e.g. in the y-direction) (see Fig. 5).
The first gradient is part of an adequate frequency encoding. It is known as the dephasing gradient. Its effect becomes obvious only due to the effect of the actual readout gradient that is applied in the first encoding direction during the spin echo after the time $T_E = 2 \cdot T_S$. Without a dephasing gradient, the nuclear spin ensemble would fan out artificially during the application of the readout gradient, thereby causing the spin echo signal to lose strength. With the dephasing gradient prior to the 180° pulse, the nuclear spin ensemble dephases already before the spin echo signal in the opposite direction of the one that is evoked by the readout gradient. After half the duration of the readout gradient ($\Leftrightarrow$ duration of the dephasing gradient), the spin echo signal has once again reached its maximum strength.

The second gradient determines the phase encoding.
Therefore, the signal information is spatially encoded in two directions in the spin echo signal. Several spin echo signals with different phase encodings enable the complete and clear spatial encoding of the two-dimensional image matrix (e.g. an x-y image matrix) (see Fig. 5).

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Theory (14/31)

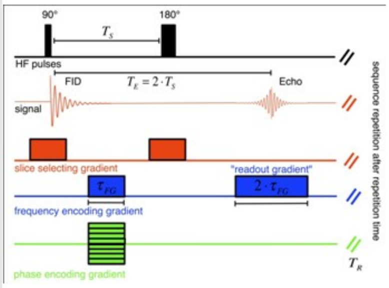

Fig. 5: Temporal application of three magnetic gradient fields for the 2D imaging of a slice. During the two HF pulses (90°, 180°), the "slice selection gradient" is applied. This gradient brings only the nuclear spins of a certain slice with a corresponding slice thickness into resonance (red). During the spin echo, the frequency encoding gradient (blue) is applied. It evokes spatial encoding in one of the two directions of the selected slice ("readout gradient"). In order to counteract a dephasing of the spins at the readout moment, an additional "dephasing gradient" that is half as long (τ_{FG}) as the readout gradient ($2\tau_{FG}$) in terms of its duration is applied prior to the 180° pulse. The spatial encoding of the second direction of the selected resonance slice is ensured by the "phase encoding gradient" (green). This gradient is applied prior to the 180° pulse. In order to obtain a real MR image with a resolution of m voxels in the phase encoding direction, the sequence that is shown must be repeated m times with different phase encoding gradients (repetition time T_R).

Theory (15/31)

Obviously, two parameters are important for 2D imaging by way of the spin echo technique. These parameters are the echo time T_E between the 90° pulse and the spin echo signal, and the repetition time T_R between consecutive measurements with different phase encoding gradients. These two parameters are used to produce different contrasts of the MR image.
The proton density contrast of an image is produced with a comparatively long T_R and a short T_E (see Fig. 6).
This becomes immediately clear, since the longitudinal magnetisations $\overrightarrow{M_L}(t)$ of different substances of the sample increase at different speeds after a first 90° HF pulse. The maximum absolute value of the longitudinal magnetisation then corresponds to the proton density of the individual substances, i.e. to the number of hydrogen protons per unit of volume. A repeated 90° HF pulse in the subsequent measurement, which is applied to the system after the repetition time T_R,converts the longitudinal magnetisation $\overrightarrow{M_L(t)}$ back into the actually detectable transverse magnetisation $\overrightarrow{M_Q(t)}$.

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Theory (16/31)

In the case of a short echo time , the strength of this transverse magnetisation is proportional to the proton density of the individual substances of the sample (proton-density-weighted image PD). For a real proton-density-weighting of the image, T_R should actually always be considerably longer than the longest relaxation time of all of the substances that are included in the sample. However, this is hardly ever practised in MR imaging because of the associated long measuring times. Common values for T_R are between 2 and 3 seconds. The echo time T_E is usually set to a value between 10 and 20 ms (see Fig. 6). Both values are based on the discernibility of oil (fat) and water (cerebrospinal fluid) as the major examination media in MR technology. The higher the proton density is in a sample substance, the brighter this substance appears in an MR image with proton-density contrast (see Fig. 6).

Theory (17/31)

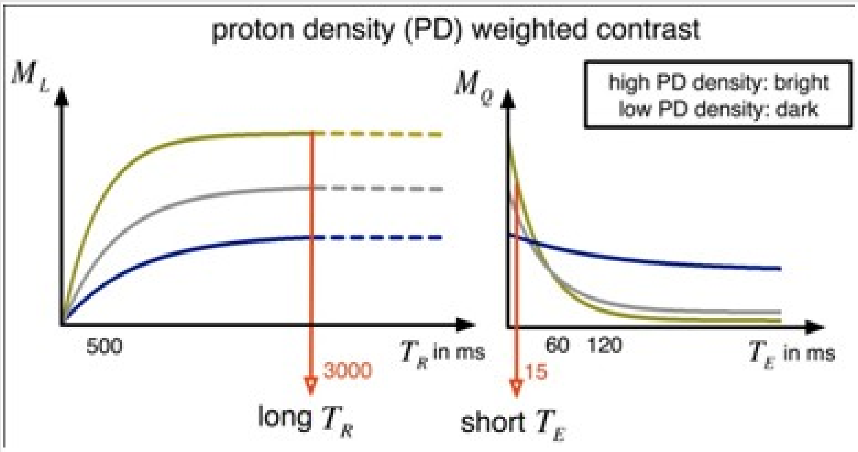

Fig. 6: Spin Echo 2D parameter settings for a proton-density-weighted MR image of a sample (PD contrast). The diagram shows the T_1 and T_2 relaxation curves for three different substances of a sample. For a PD contrast, a comparatively long repetition time T_R must be selected (e.g. 3000 ms). The echo time T_E must be comparatively short (e.g. 15 ms). In the case of the PD contrast, substances with a high hydrogen proton density appear brighter and substances with a low hydrogen proton density appear darker.

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Theory (18/31)

The T_2 contrast of an image is produced with a comparatively long T_E and a long T_R (see Fig. 7). If, after the repeated 90° pulse in the subsequent measurement (see PD contrast), a certain echo time T_E passes before the spin echo signal is detected, the transverse magnetisation $\overrightarrow{M_Q(t)}$ will have decayed already to a rather large extent. This occurs within the relaxation time T_2 that is characteristic for the individual substances of the sample. The T_2 relaxation curves of the individual substances of the sample cross each other and the influence of the proton density gets lost (see Fig. 7). As a result, the signal strength of the spin echo mainly depends only on the T_2 decay. For a T_2 contrast of an MR image, repetition times T_R between 2 and 3 seconds and echo times between 60 and 120 ms are usually used. Both values are based on the discernibility of oil (fat) and water (cerebrospinal fluid) as the major examination media in MR technology. In the T_2-weighted MR image, water (cerebrospinal fluid) with a long T_2 appears bright and oil (fat) with a short T_2 appears considerably darker (see Fig. 7).

Theory (19/31)

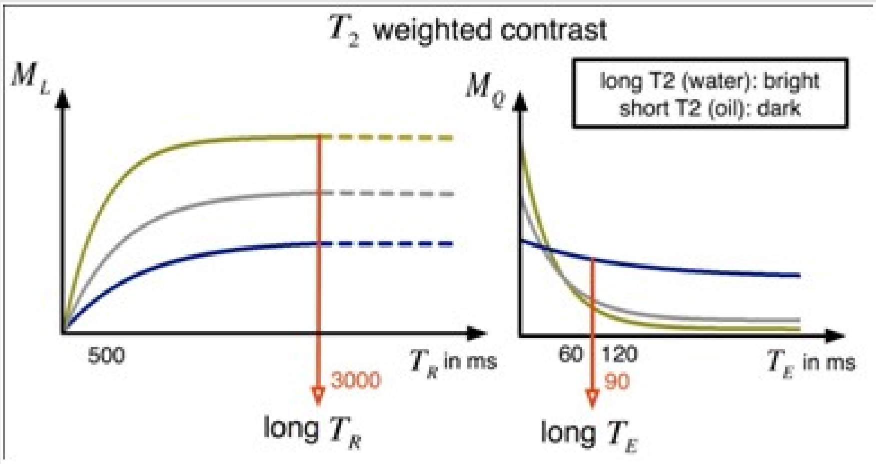

Fig. 7: Spin Echo 2D parameter settings for a T_2-weighted MR image of a sample (T_2 contrast). The diagram shows the T_1 and T_2 relaxation curves for three different substances of a sample. For a T_2 contrast, a comparatively long repetition time T_R must be selected (e.g. 3000 ms). The echo time T_E must also be comparatively long (e.g. 90 ms). In the case of the T_2 contrast, substances with a long T_2 relaxation time (e.g. water) appear bright and substances with a short T_2 relaxation time (e.g. oil) appear dark.

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Theory (20/31)

The T_1 contrast of an image is produced with a comparatively short T_R and a short T_E (see Fig. 8). A short T_R directly implies a weaker signal, since the longitudinal magnetisation or the state of equilibrium has not been completely restored before the repeated measurement sequence commences. Correspondingly, the contrasts would also decrease with an increasing T_E,which is why a relatively short T_E must be selected (see Fig. 8). This also suppresses the influence of the T_2 relaxation. Consequently, the recorded signal mainly results from the restored longitudinal magnetisation after the time T_R.Therefore, it is directly proportional to the T_1 relaxation times of the individual substances of the sample. For a T_1 contrast of an MR image, repetition times T_R between 250 and 500 ms and echo times between 10 and 20 ms are usually used. Both values are based on the discernibility of oil (fat) and water (cerebrospinal fluid) as the major examination media in MR technology. In the- T_1 weighted MR image, water (cerebrospinal fluid) with a long T_1 appears dark and oil (fat) with a short T_1 appears considerably brighter (see Fig. 8).

Theory (21/31)

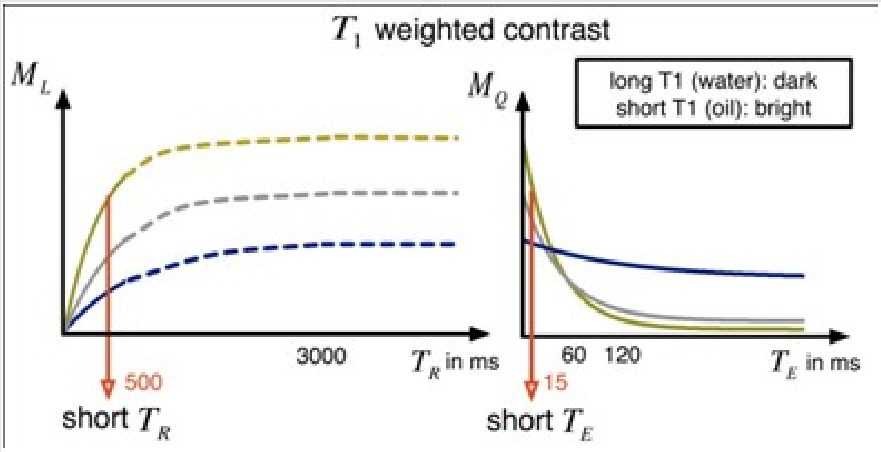

Fig. 8: Spin Echo 2D parameter settings for a T_1-weighted MR image of a sample (T_1 contrast). The diagram shows the T_1 and T_2 relaxation curves for three different substances of a sample. For a T_1 contrast, a comparatively short repetition time T_R must be selected (e.g. 500 ms). The echo time T_E must also be comparatively short (e.g. 15 ms). In the case of the T_1 contrast, substances with a long T_1 relaxation time (e.g. water) appear dark and substances with a short T_1 relaxation time (e.g. oil) appear bright.

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Theory (22/31)

Depending on the set parameters, 2D Spin Echo imaging provides highly precise MR images, but it is also associated with a longer measuring time, since the repetition times T_R must be comparatively long (see above). The question is whether the measuring times can be reduced by way of a specific measurement sequence or special settings. In 1985, such a method for rapid MR imaging was developed by A. Haase, J. Frahm, D. Matthaei, W. Hanicke, and K.-D. Merboldt at the Max-Planck-Institut für biophysikalische Chemie in Göttingen (Germany). It is called FLASH imaging.
The idea behind FLASH imaging is the initial deflection of the magnetisation vector by an angle that is considerably smaller than 90°. In the case of smaller angles, only a part of the original longitudinal magnetisation $\overrightarrow{M_{L0}}$ is converted into a transverse magnetisation $\overrightarrow{M_Q(t)}$. On the other hand, this means that a certain proportion of a longitudinal magnetisation is always included in the total magnetisation vector and that $\overrightarrow{M_L(t)}$ is unequal to zero at all times. The smaller transverse magnetisation proportion $\overrightarrow{M_Q(t)}$ evokes a weaker signal that can be measured nevertheless (see Fig. 0).

Theory (23/31)

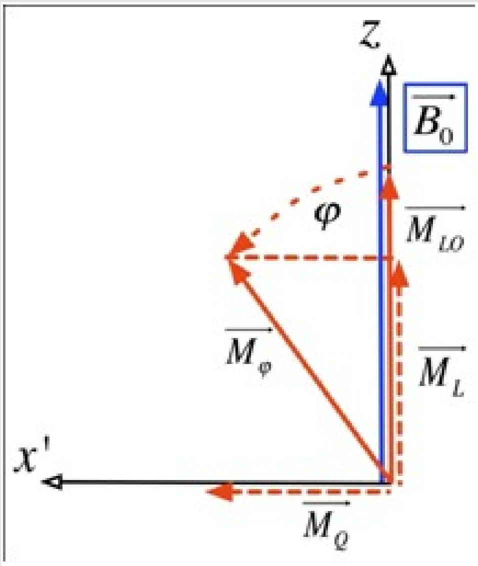

Fig. 9: Deflection of the original magnetisation vector $\overrightarrow{M_L}$ by an angle ϕ.

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Theory (24/31)

A 20° pulse, for example, produces a transverse magnetisation of 34% of the maximum value, while the remaining longitudinal magnetisation still corresponds to 94% of the maximum value. As a result, the longitudinal magnetisation proportion is already rather high at the beginning of a new measurement sequence (due to the relaxation, it is even higher than 94%).
This means that a 20° pulse can generate a stronger MR signal than a 90° pulse if the repetition times $T_R < T_1$ are very short. For a given repetition time T_R, there is a well-defined flip angle at which the signal of a specific substance with a characteristic T_1 becomes maximal. This flip angle is known as the ERNST angle. However, due to the exponential growth process, the relaxation in the case of small deflection angles is slower than in the case of large deflection angles. As a result, there are two opposite tendencies concerning the deflection of the magnetisation vector by a certain angle. The smaller the deflection angle is the stronger the longitudinal magnetisation is in consecutive measurement sequences. At the same time, the relaxation back to the ground state takes more time. The setting of the deflection angle at which the longitudinal magnetisation remains identical after every new measurement sequence for a certain substance (state of equilibrium) is called STEADY STATE setting.

Theory (25/31)

In most cases, the FLASH method uses a so-called gradient echo instead of a spin echo as the measurement signal. Gradient echoes can be used for the direct restoration of an artificially dephased FID signal (free induction decay). In "Fundamental principles of nuclear magnetic resonance (NMR)", we have already seen that the FID signal typically decays in a time $T_2^* < T_2$. However, if a gradient of the duration τ_{FD} is applied directly after the HF pulse that generates the FID signal, i.e. still within the time T_2^*, the nuclear spins dephase much more quickly and the measurement signal gets lost in a time $T_2^{**} < T_2^* < T_2$.
Caused by a gradient with reversed polarity and the duration $2 \cdot \tau_{FD}$, the spins rephase, and we measure an echo during the restoration of the FID (gradient echo). The amplitude of the gradient echo is not proportional to the relaxation time T_2, as was the case with the spin echo. Instead, it is proportional to the relaxation time .T_2^* Of course, the gradient echo also involves a characteristic echo time T_E, i.e. the time between the HF pulse and the maximum of the gradient echo. It can be considerably shorter than in the case of the spin echo. In addition, the gradient echo is still relatively strong in the case of small excitation angles, which means that it supplies an adequate measurement signal (better SNR per unit of time).

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Theory (26/31)

Smaller excitation angles also mean that the repetition time T_R can be strongly reduced, which leads to additional time savings. The disadvantages compared to the spin echo are the stronger influence of magnetic susceptibilities and inhomogeneities of the external static magnetic field $\vec{B_0}$. This results in a poorer image quality of flash MR images.

We will now once again summarise the described procedure for magnetic resonance imaging based on the gradient echo technique (Flash 2D). In doing so, we will put special emphasis on the sequence that is used as well as on the most important parameters (see Fig. 10).

The starting point for our considerations is the state of equilibrium, i.e. the alignment of the magnetisation vector parallel to the static magnetic field $\vec{B_0}$. An HF pulse, which fulfils the resonance condition (1), deflects the magnetisation vector by a certain angle ϕ. ϕ depends on the duration of the HF pulse (see "Fundamental principles in nuclear magnetic resonance (NMR)"). If the HF pulse is additionally superimposed by a magnetic gradient field, ϕ can also be adjusted via the amplitude of the gradient (this is useful in some cases in order to generate images that are more robust in view of inhomogeneities; see destructive interference).

Theory (27/31)

The tilted magnetisation vector continues to precess around $\vec{B_0}$ and the nuclear spin ensemble would dephase in a time $T_2^* < T_2$.(normal FID signal) if no influence was exerted. However, shortly after the exciting HF pulse (variation possible via the echo time T_E), a gradient of the duration τ_{FD} is applied. This gradient artificially dephases the normal FID signal and, thereby, destroys the FID after half of the gradient duration. After the time $T_2^{**} < T_2^*$. an FID signal can no longer be detected. At the same time as this dephasing gradient, the phase encoding gradient, which has already been explained in detail, is applied. It enforces spatial encoding in one of the two directions of the 2D image matrix. Only the repeated application of this phase encoding gradient with different amplitudes in consecutive measurement sequences of the measurement cycle can lead to a complete, resolution-dependent spatial encoding (see Fig. 10).

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Theory (28/31)

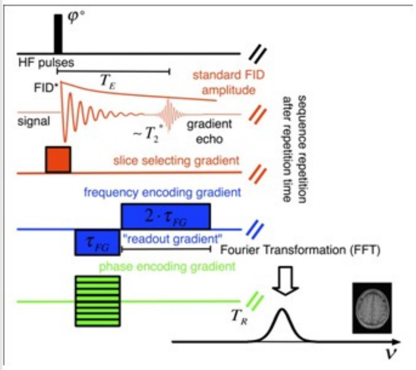

Fig. 10: Recording of a 2D MR image by way of a gradient echo (Flash 2D). The slice selection gradient is applied during the exciting HF pulse ($\phi°$). This gradient brings only the nuclear spins of a certain slice with a corresponding slice thickness into resonance (red). A dephasing gradient (blue) with the duration τ_{FG} dephases the nuclear spins during the FID signals. As a result, the FID signal is lost. Immediately thereafter, the FID signal is restored by way of a gradient with the duration $2 \cdot \tau_{FG}$ and with opposite polarity (gradient echo, blue). This gradient is also used as the frequency encoding gradient. It evokes spatial encoding in one of the two directions of the selected slice ("readout gradient"). Spatial encoding of the second direction of the selected resonance slice is ensured by the phase encoding gradient (green). It is applied simultaneously with the above-mentioned dephasing gradient. In order to obtain a real MR image with a resolution of m voxels in the phase encoding direction, the sequence that is shown must be repeated m times with different phase encoding gradients (repetition time T_R).

Theory (29/31)

Directly after the phase encoding gradient and dephasing gradient, the frequency encoding gradient with the duration $2 \cdot \tau_{FD}$ is applied. It has the opposite polarity of the dephasing gradient and brings the individual phases of the nuclear spins back into phase so that a maximum signal is once again generated after half the duration of the frequency encoding gradient. This signal is the gradient echo. It restores the decayed FID signal. As a result, its amplitude is proportional to the relaxation time T_2^+ (see Fig. 10). The gradient echo is now frequency-encoded and phase-encoded at the same time. As a result, it supplies all of the necessary information for an adequate reconstruction of the MR image. Once again, this reconstruction is realised by way of a Fourier transformation. The Spin Echo 2D method and the Flash 2D method are actually rather similar, except for the fact that Flash 2D imaging uses considerably shorter repetition times T_R and also shorter echo times T_E and MR images can be acquired after a comparatively short measuring time.

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Theory (30/31)

Obviously, three parameters are important for 2D imaging by way of the gradient echo technique (FLASH). These parameters are the echo time T_E between the 90° pulse and the gradient echo signal, the excitation angle ϕ, and the repetition time T_R between consecutive measurements with different phase encoding gradients. These three parameters are used to produce different contrasts of the MR image.
The T_1 contrast is produced with relatively short echo and repetition times. The excitation angle ϕ, on the other hand, must be comparatively large (e.g. 40 ms < T_R < 150 ms, 3 ms < T_E < 10 ms, 30°< ϕ < 80° in order to be able to distinguish water (cerebrospinal fluid) clearly from oil (fat)).
The T_2^* contrast is produced with relatively long echo and repetition times. The excitation angle ϕ, on the other hand, must be comparatively small (e.g. T_R approx. 500 ms, 18 ms < T_E < 40 ms, 5°< ϕ < 20° in order to be able to distinguish of water (cerebrospinal fluid) clearly from oil (fat)).
The proton-density contrast is produced with relatively long repetition times, but short echo times. The excitation angle ϕ must be comparatively small (e.g. T_R approx. 500 ms, 3 ms < T_E < 10 ms, 5°< ϕ < 20° in order to be able to distinguish water (cerebrospinal fluid) clearly from oil (fat)).

Theory (31/31)

Under certain circumstances and if the contrast methods are selected accordingly, the repetition time T_R and the echo time T_E can be even shorter (the values stated above are guidance values). The special parameter settings for the generation of all of the Flash 2D contrasts that are mentioned above become directly clear based on the considerations concerning the Spin Echo 2D method.
The described parameter settings for a specific contrast are highly system-dependent, since the Flash 2D method strongly depends on the magnetic homogeneity of the magnetic field $\overrightarrow{B_0}$. In systems with a low level of homogeneity, an even shorter echo time than the one described should be selected in order to ensure a good image quality. The repetition time can also be further reduced in the case of smaller excitation angles.
The Flash 2D imaging method can be made even quicker (FastFlash 2D) by optimising the parameters. However, this can only be realised at the expense of the SNR and resolution. For example, it is possible to fix the number of data points and phase steps on a relatively low value. This also reduces the scanning window and enables even shorter echo times. Please refer to the corresponding experiment in the course "Imaging 1" of the "measure MRT" software in order to intensify your understanding of FastFlash 2D imaging.

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Equipment

Position	Material	Item No.	Quantity
1	PHYWE Compact magnetic resonance tomograph (MRT)	09500-99	1

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Setup and Procedure

Setup (1/2)

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Set the MR unit up as shown in Fig. 11. Ensure that the unit is used in a dry and dust-free room. Ensure that the unit is set up in a vibration-free manner. The mains power switch and the device connector must be freely accessible. Ensure that the ventilation slots are not blocked or covered. Keep a suitable safety distance from other technical equipment and storage media, since they may be damaged by strong magnets. Remove any metallic objects in the direct vicinity of the unit. Ensure that the POWER switch of the control unit is set to off (see Fig. 13). Connect the control unit via the power supply connector (12 V DC, 2 A) to the power supply. It is absolutely necessary to use the power supply unit that is intended for this purpose (see Fig. 13).

Fig. 11: Set-up of the MRT training unit

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Setup (2/2)

Connect the control unit and the magnet by way of the gradient and BNC cables that are intended for this purpose (see Fig. 12). Then, connect the USB interfaces of the control unit and measurement computer via a USB 2.0 high-speed cable (see Fig. 13). Switch the unit on via the POWER rocker switch (the MR unit should only be switched on for performing experiments). When the unit is started for the first time, the operating system of the computer will recognise the control unit. Then, install the device driver and measurement software (see the installation instructions). Start the "measure MRT" software.

Fig. 12: Magnet and control unit connectors

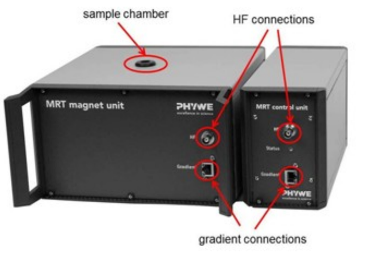

Fig. 13: Connectors at the back of the control unit

Procedure (1/9)

When the "measure MRT" software is started, a window will open automatically as shown in Fig. 14. In area 1, experiments can be selected (experiments area). The associated parameters are displayed in area 2 (parameters area). Area 3 shows a sequence representation of the selected experiment (sequence area). Finally, the results are displayed in area 4 (results area). All of these areas can be arranged as desired in the window. An individual arrangement can be saved for future measurements via the "program settings".

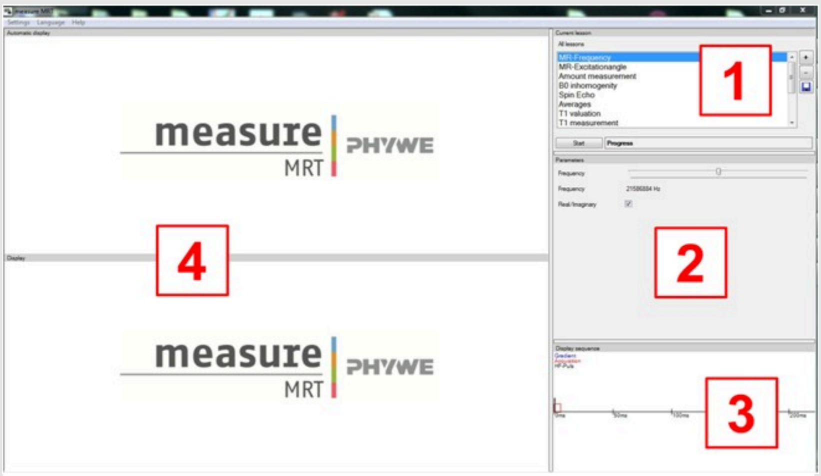

Fig. 14: Areas of the "measure MRT" program

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Procedure (2/9)

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Note:

The following experiments (A-C) should be performed in chronological order. In order to be able to do this, the experiment ensembles "Fundamental principles of nuclear magnetic resonance" (P5942100) and "Relaxation times in nuclear magnetic resonance" (P5942200) should have been performed. In this case, the settings for these ensembles have been saved for the experiment ensemble "Spatial encoding in nuclear magnetic resonance". It is always useful to repeat the experiments concerning the MR frequency and MR excitation angle, to attune the system frequency once more precisely to the Larmor frequency, and also to adjust the pulse duration of an ideal 90° HF pulse (compare P5942100). This is why these experiments are integrated in all of the courses that include the following experiment ensemble. TIP: Prior to starting the experiment, check the parameter settings of the experiment ensemble "Fundamental principles of nuclear magnetic resonance" (P5942100) and readjust them, if necessary.

Procedure (3/9)

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A: Application of frequency and phase encoding for the generation of a spin echo signal based on which a 2D image can be acquired (Spin Echo 2D)

1. Place the 5 mm water sample and 5 mm oil sample into the sample chamber of the MR unit (double sample) at the same time. Align the double sample so that the two samples are aligned on one line and parallel to the rear edge of the magnet housing. In the experiments area (lessons), select the lesson Spin echo 2D. The parameters area shows the setting options Data points, Phase steps, Slice orientation, Read gradient, Phase gradient, Repetition time, Averages, and Echo time (see Fig. 15). Set the Repetition time to 300 ms, the number of Averages to 5, and the Echo time to 15 ms. The Read and Phase gradient should be set to approximately 70 μTs/m. Then, set the number of image points per echo (Data points) to 128, the number of spatially encoded echoes (Phase steps) to 64, and select the X-Z slice (Slice orientation) as the plane of the 2D slice through the sample. Start the measurement.

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Procedure (4/9)

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2. Maintain the settings of 1. Change the plane of the 2D slice through the sample by way of the Slice orientation setting. Then, acquire a 2D MR image of the X-Y slice (coronal slice) and of the Y-Z slice (sagittal slice).

3. Select the X-Z slice once again via the Slice orientation setting (transverse slice). Vary the number of image points per echo by way of the Data points slider and the number of phase steps by way of the Phase steps slider. Maintain the other settings of 1. Record various images with different numbers of data points and phase steps.

4. Once again, use the settings of 1, i.e. set the Data points to 128 and the Phase steps to 64. Vary the strength of the readout and phase gradients by way of the sliders Read gradient and Phase gradient and record the corresponding transverse MR images.

Procedure (5/9)

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5. Once again, use the settings of 1, i.e. set the Read gradient and Phase gradient to approximately 70 μTs/m. Vary the repetition time and echo time by way of the sliders Repetition time and Echo time and record the corresponding transverse MR images.

6. Once again, use the settings of 1, i.e. set the repetition time to 300 ms and the echo time to 15 ms. Vary the number of averages by way of the slider Averages and record the corresponding transverse MR images.

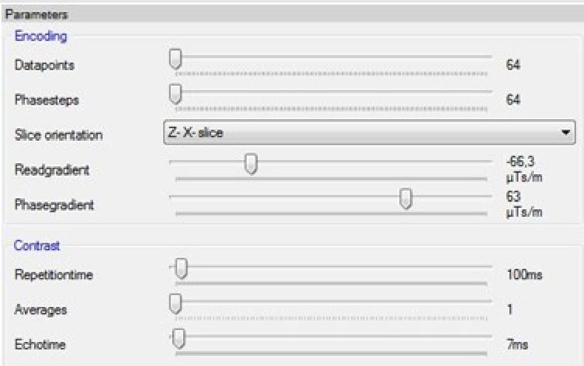

Fig. 15: Spin echo 2D - parameters

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Procedure (6/9)

7. Replace the double sample of oil and water with the structure sample in the MR sample chamber. Use the settings of 1 and select an X-Y slice, X-Z slice, and Y-Z slice via the setting Slice orientation. Record the corresponding MR images (coronal, transverse, and sagittal). Compare the images and draw conclusions concerning the structure of the plastic material in the structure sample.

Procedure (7/9)

B: Application of frequency and phase encoding for the generation of a gradient echo signal based on which a 2D image can be acquired (Flash 2D)

1. Place the 5 mm water sample and 5 mm oil sample into the sample chamber of the MR unit (double sample) at the same time. Align the double sample so that the two samples are aligned on one line and parallel to the rear edge of the magnet housing. In the experiments area (lessons), select the lesson Flash 2D. The parameters area shows the setting options Data points, Phase steps, Slice orientation, Read gradient, Phase gradient, Repetition time, Echo time, Averages, and Angle (see Fig. 16). Set the Repetition time to 40 ms, the number of Averages to 10, the Echo time to 5 ms, and the Angle to 30°. The Read and Phase gradient should be set to approximately 70 μTs/m. Then, set the number of image points per echo (Data points) to 128, the number of spatially encoded echoes (Phase steps) to 64, and select the X-Z slice (Slice orientation) as the plane of the 2D slice through the sample. Start the measurement.

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Procedure (8/9)

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2. Use the settings of 1 again. Vary the repetition time and echo time by way of the sliders Repetition time and Echo time and record the corresponding transverse MR images.

3. Use the settings of 1 again. Vary the excitation angle by way of the slider Angle and record the corresponding transverse MR images.

4. Replace the double sample of oil and water with the structure sample in the MR sample chamber. Set the Repetition time to 30 ms, the number of Averages to 30, the Echo time to 3 ms, and the Angle to 20°. The Read and Phase gradient should be set to approximately 70 μTs/m. Set the number of image points per echo (Data points) to 128 and the number of spatially encoded echoes (Phase steps) to 64. Select an X-Y, X-Z, and Y-Z slice by way of the setting Slice orientation. Record the corresponding MR images (coronal, transverse, and sagittal). Compare the images and draw conclusions concerning the structure of the plastic material in the structure sample.

Procedure (9/9)

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Fig. 16: Flash 2D - parameters

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Evaluation

A: 2D spatial encoding by frequency encoding (1/19)

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Record a transverse 2D MR image of a double sample of oil and water by way of the spin echo method. In a transverse 2D MR image, the cross-sectional plane is parallel to the bottom of the magnetic resonance tomography scanner.

Fig. 17 shows a transverse MR image of the double sample of oil and water that is aligned in parallel to the rear edge of the magnet housing. The image has been recorded by way of a spatially encoded spin echo signal (Spin Echo 2D). The following settings were used for the reconstruction: Repetition time: 300 ms; echo time: 15 ms; number of averages: 5; readout gradient: 70.2 μTs/m phase gradient: 70.2 μTs/m (variable in consecutive measurement sequences of an averaging step); number of image points per echo (Data points): 128; number of spatially encoded echoes (Phase steps): 64; plane of the 2D slice: X-Z slice (Slice orientation). In accordance with the theory, the recorded MR image provides a T_1 contrast, since the repetition time T_R and the echo time T_E are comparatively short. In this case, oil appears bright and water dark, i.e. the two substances of the double sample can be clearly identified. In Fig. 17, the sample on the left is the water sample and the sample on the right is the oil sample.

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A: 2D spatial encoding by frequency encoding (2/19)

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Please note, however, that the recorded image is a summation of the signals over the entire sample space, since an explicit slice selection with a well-defined thickness was not specified. Directly under the transverse MR image of the double sample in the position space in Fig. 17, the image of this sample in the k-space or wave-vector space is shown. The data points in the centre of the k-space determine the signal-to-noise ratio, structure, and contrast of the reconstructed image. The outer data points provide information concerning borders, edges, contours, and fine transitions.

A: 2D spatial encoding by frequency encoding (3/19)

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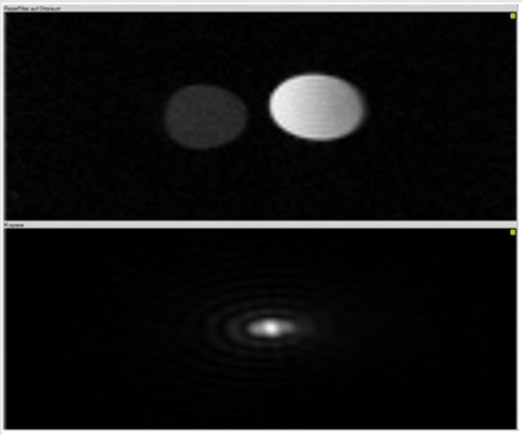

Fig. 17: Transverse (X-Z) MR image of the double sample of oil and water with the T_1 contrast. The double sample is aligned in parallel to the rear edge of the magnet housing. The image has been recorded by way of a spin echo signal (Spin Echo 2D). For this purpose, the repetition time was set to 300 ms and the echo time to 15 ms. 5 individual images were used for averaging. The two individual samples can be clearly identified in the position space (top), since oil appears bright in the T_1 contrast and water appears dark. In the wave-vector space (k-space), the data points in the centre of the k-space determine the signal-to-noise ratio, structure, and contrast in the reconstructed image. The outer data points provide information concerning borders, edges, contours, and fine transitions. The conversions between the position space and k-space are realised by way of Fourier transformations.

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A: 2D spatial encoding by frequency encoding (4/19)

Generate 2D MR images of the double sample with the settings of 1 for all of the three slice orientations that can be selected (X-Y, X-Z, Y-Z). Please note that the generated images are accumulated images over the entire sample space, which means that the selected slice always has the thickness of the entire sample space.

Figs. 18 a and b show the MR image of the double sample of oil and water for the X-Y slice (a) and for the Y-Z slice (b). The parameter settings are identical with the ones of task 1. This means that these two MR images have been recorded with the T_1 contrast. Fig. 18 a obviously shows a joint signal of the double sample, i.e. the individual samples cannot be distinguished. This is directly due to the special position of the double sample. In the X-Y slice, both samples are located directly one behind the other in the sample chamber. Since the image corresponds to a summation of all of the signals of the sample space, the signals of the oil and water samples coincide. Upon a closer look, however, it becomes clear that actually two different samples were analysed. This results from small shifts of the two individual samples with regard to one another, since the exact positioning of the double sample parallel to the rear edge of the magnet housing is hardly possible.

A: 2D spatial encoding by frequency encoding (5/19)

In Fig. 18 b, the oil sample and the water sample can be distinguished once again, since in the Y-Z slice the two samples are located next to each other. With the recorded T_1 contrast, it is clear that the sample on the left is the water sample and the sample on the right the oil sample.

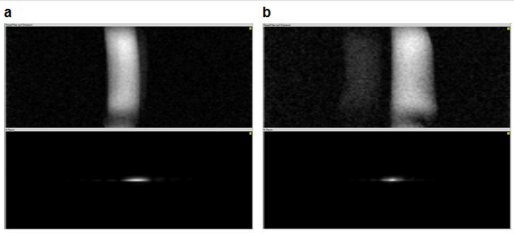

Fig. 18: Coronal (X-Y) (a) and sagittal (Y-Z) (b) MR images of the double sample of oil and water with the T_1 contrast. The double sample is aligned in parallel to the rear edge of the magnet housing.

The images have been recorded by way of a spin echo signal (Spin Echo 2D). For this purpose, the repetition time was set to 300 ms and the echo time to 15 ms. 5 individual images were used for averaging. In a, the two individual samples of the double sample cannot be distinguished, since the signals of the oil and water sample coincide in the X-Y slice. In b, the samples can be clearly identified. The sample on the left is the water sample and the sample on the right the oil sample. For both MR images, the corresponding representation in the k-space is also shown (bottom).

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A: 2D spatial encoding by frequency encoding (6/19)

Study the effects of the number of data points and phase steps on the transverse 2D MR image of the double sample.

It is clear that the number of data points and phase steps determines the resolution of the MR image. The higher the number of data points and phase steps is the more grey values that must be assigned. As a result, the MR image becomes finer and more detailed. The initial resolution of 128 data points x 64 phase steps means that 128 different grey values must be assigned to the 128 voxels in the x-direction, and 64 different grey values to the 64 voxels in the y-direction. This is analogous to the resolution of display screens.
Of course, an image quality that is as high as possible and also a highly detailed image are desirable, which means that the number of data points and phase steps must be as high as possible. However, this is when the "imaging process duration" factor comes into play. Based on our considerations concerning the Spin Echo 2D method (see the theory), a higher number of data points does not require more time, since all of the data points of a single measurement sequence of the measurement cycle can be written into the corresponding line of the data matrix. The situation is different as far as the number of phase steps is concerned.

A: 2D spatial encoding by frequency encoding (7/19)

Every phase step requires a new measurement sequence because, for every measurement sequence, only one specific phase position can be written into a column of the data matrix. This means that if the number of phase steps is doubled, the recording time of the MR images also doubles. As a result, it is the number of phase steps that determines nearly entirely the duration of an MR imaging process (compare the duration of the MR imaging processes in medical diagnosis applications).
Figs. 19 a and b show the transverse MR image of the double sample of oil and water with two different resolutions. In Fig. 19 a, the number of data points was increased to 256 compared to Fig. 17, while the number of phase steps (64) was maintained. Regardless of the higher number of data points in the x-direction, the time that is required for this imaging process is identical with the time that is required in the case of fewer data points. In Fig. 19 b, the number of phase steps was increased to 128 compared to Fig. 17, while the number of data points (128) was maintained. As a result, the time that was required for the imaging process with 128 phase steps in the y-direction was twice as long as the time that was required for the imaging process with 64 phase steps in the y-direction.

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A: 2D spatial encoding by frequency encoding (8/19)

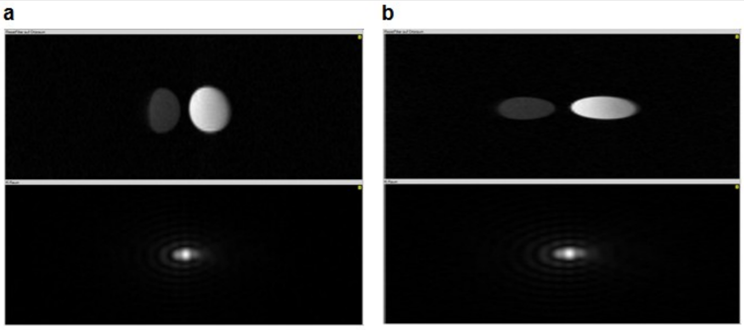

Fig. 19: Transverse (X-Z) MR images of the double sample of oil and water with the T_1 contrast. The double sample is aligned in parallel to the rear edge of the magnet housing. The images have been recorded by way of a spin echo signal (Spin Echo 2D). For this purpose, the repetition time was set to 300 ms and the echo time to 15 ms. 5 individual images were used for averaging. In a, the MR image was recorded with 256 data points in the x-direction and 64 phase steps in the y-direction. In b, the MR image was recorded with 128 data points in the x-direction and 128 phase steps in the y-direction.

A: 2D spatial encoding by frequency encoding (9/19)

Record various transverse 2D MR images of the double sample with different strengths of the readout and phase gradients.

The spatially encoded representation of the MR image can be varied by way of the settings concerning the strength of the readout and phase gradients. The steeper a magnetic gradient field is the greater the frequency differences are with which the nuclear spins precess in the individual voxels (see the resonance condition). In accordance with the Fourier transformation, this automatically leads to a higher or wider spatially encoded MR image. Of course, the magnetic gradient field can also be simply reversed. In this case, the nuclear spins that have precessed with the highest frequency in certain voxels now precess with the lowest frequency in these voxels. After a Fourier transformation, this effectively leads to a mirroring of the MR image in the position space.

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A: 2D spatial encoding by frequency encoding (10/19)

Figs. 20 a-d verify our considerations for the transverse MR image of the double sample of oil and water. With the exception of the gradient strength, the settings of task 1 were adopted (a: readout gradient: 130.0 μTs/m, phase gradient: 70.2 μTs/m; b: readout gradient: 40.0 μTs/m, phase gradient: 70.2 μTs/m; c: readout gradient: 27.3, μTs/m; phase gradient: 27.3 μTs/m; d: readout gradient: -70.2 μTs/m, phase gradient: -70.2 μTs/m). Due to the reversal of the magnetic gradient field in Fig. 20 d compared to Fig. 17, the double sample is now also reversed in the spatially encoded representation. This means that the oil sample is on the left and the water sample on the right.

A: 2D spatial encoding by frequency encoding (11/19)

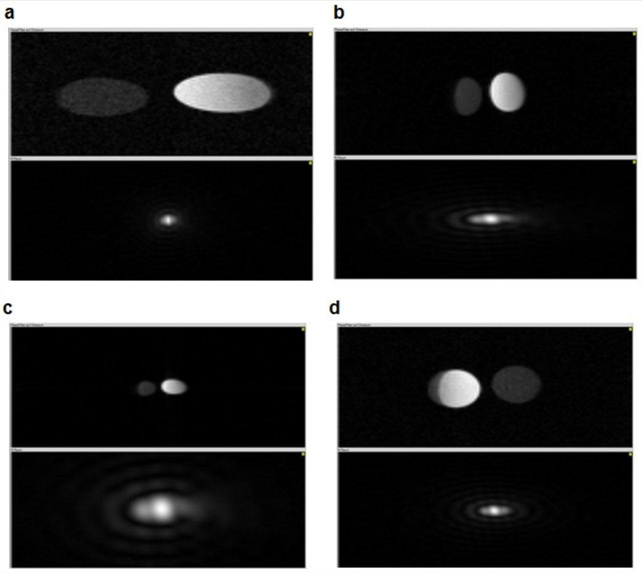

Fig. 20: Transverse (X-Z) MR images of the double sample of oil and water with the T_1 contrast. The double sample is aligned in parallel to the rear edge of the magnet housing. The images have been recorded by way of a spin echo signal (Spin Echo 2D). For this purpose, the repetition time was set to 300 ms and the echo time to 15 ms. 5 individual images were used for averaging. The following strength values of the gradient fields were used for the image acquisition. a: readout gradient: 130.0 μTs/m, phase gradient: 70.2 μTs/m ; b: readout gradient: 40.0 μTs/m, phase gradient: 70.2 μTs/m; c: readout gradient: 27.3 μTs/m, phase gradient: 27.3 μTs/m; d: readout gradient: -70.2 μTs/m, phase gradient: -70.2 μTs/m.

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A: 2D spatial encoding by frequency encoding (12/19)

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Record various transverse 2D MR images of the double sample with different repetition and echo times.

The influence of the repetition and echo time on the contrast of an MR image has already been described in detail in the theory part. Depending on the setting of these two parameters, a T_1 contrast, T_2 contrast, or proton-density contrast (PD contrast) can be generated. For a PD contrast, the repetition time T_R must be comparatively long and the echo time T_E must be comparatively short. For a T_2 contrast, the repetition time T_R and the echo time T_E must be comparatively long. For a T_1 contrast, on the other hand, the repetition time T_R as well as the echo time T_E must be comparatively short. As a result, it is directly clear that only the T_2 contrast can be generated relatively quickly.
In medical diagnostics, the contrast setting that can be expected to produce the highest contrast between the medium to be examined and the surrounding substance is usually selected T_2, for example, provides a strong contrast between liquid and solid or semi-solid substances and the T_2 contrast is particularly suitable for distinguishing fluid accumulations or some types of tumours from the surrounding tissue.

A: 2D spatial encoding by frequency encoding (13/19)

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Figs. 21 a-c show the transverse 2D MR images of the double sample of oil and water with the three contrasts that are mentioned above. With the exception of the repetition and echo times, the settings of task 1 were adopted. In Fig. 21 a, the PD contrast was generated with a repetition time of 3000 ms and an echo time of 15 ms (see the theory). Since the oil and water samples have approximately the same level of brightness, it can be concluded that the hydrogen proton densities of oil and water in the two 5 mm samples are relatively similar. In Fig. 21 b, the T_2 contrast was generated with a repetition time of 3000 ms and an echo time of 90 ms (see the theory). With the T_2 contrast, the water sample appears brighter than the oil sample, since water has a considerably longer T_2 relaxation time. As a result, the samples can be clearly identified in this picture. The sample on the left is the water sample and the sample on the right the oil sample. In Fig. 21 c, the T_2 contrast was generated with a repetition time of 500 ms and an echo time of 15 ms (see the theory and task 1). With the T_1 contrast, the oil sample appears brighter than the water sample, since oil has a considerably shorter T_1 relaxation time. As a result, the sample can also be clearly identified with the T_1 contrast.

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A: 2D spatial encoding by frequency encoding (14/19)

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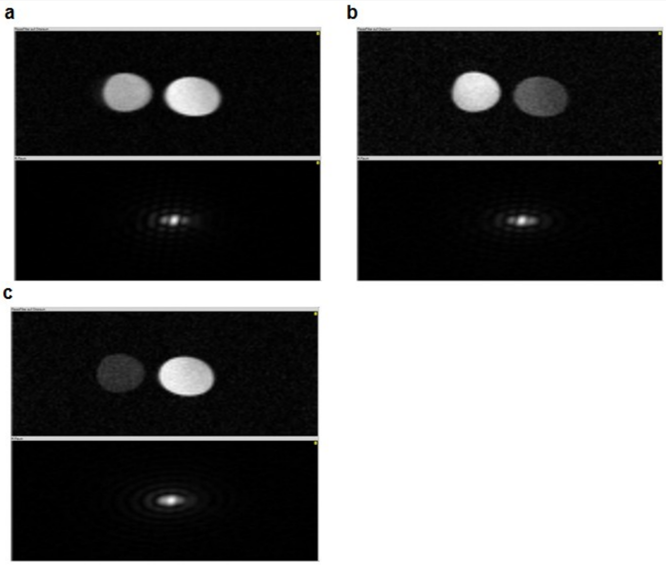

Fig. 21: Transverse (X-Z) MR images of the double sample of oil and water with three different contrasts. The double sample is aligned in parallel to the rear edge of the magnet housing. The images have been recorded by way of a spin echo signal (Spin Echo 2D). 5 individual images were used for averaging. In a, the MR image of the double sample has been generated with the proton-density contrast (PD) (repetition time: 3000 ms; echo time: 15 ms). In b, the MR image of the double sample has been generated with the T_2 contrast (repetition time: 3000 ms; echo time: 90 ms). In c, the MR image of the double sample has been generated with the T_1 contrast (repetition time: 500 ms; echo time: 15 ms).

A: 2D spatial encoding by frequency encoding (15/19)

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Study the influence of the number of averages on the transverse 2D MR image of the double sample of oil and water.

In MR technology, the number of averages is an important means for drastically increasing the quality of the MR images. During the averaging process, several "identical" measurements are performed consecutively. These measurements are then used for averaging. The random noises of the individual measurements cancel each other partly out, while the included MR signals add up. This method makes use of the fact that the MR signal is identical in every individual measurement, whereas the included noise is random. The key is, however, that the improvement of the signal-to-noise ratio (SNR) is proportional to the root of the number of averages. This means that four averages are needed in order to double the signal-to-noise ratio. Averaging is very time-consuming but often indispensable for a good image quality. A particularly high number of averages is needed when the MR signal is comparatively weak, i.e. if only one single thin slice of a sample is scanned. With the method that has been described up to now, we always detect a signal that has been accumulated (added up) over the entire sample space. With such an accumulated signal, reliable MR images can already be produced with a relatively small number of averages.

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A: 2D spatial encoding by frequency encoding (16/19)

Figs. 22 a and b show the transverse MR images of the double sample of oil and water that were produced with "1 average" (a) and 10 averages (b). The settings of task 1 were maintained. The double sample can be clearly identified in Fig. 22 a in spite of a smaller SNR. However, the SNR improvement is clearly recognisable in Fig. 22 b. The measuring time for the MR image in Fig. 22 a was 19.2 s and the measuring time for the MR image in Fig. 22 b was 192 s.

A: 2D spatial encoding by frequency encoding (17/19)

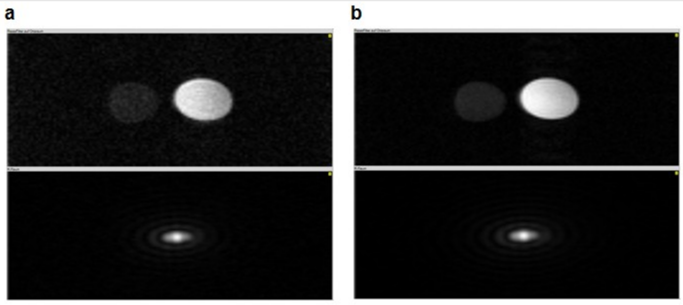

Fig. 22: Transverse (X-Z) MR images of the double sample of oil and water with the T_1 contrast. The double sample is aligned in parallel to the rear edge of the magnet housing. The images have been recorded by way of a spin echo signal (Spin Echo 2D). For this purpose, the repetition time was set to 300 ms and the echo time to 15 ms. In a, only one single transverse MR image was recorded, whereas the average of 10 consecutive measurements was taken for the image in b. The result is that the MR image of the double sample in b in much sharper than in a.

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A: 2D spatial encoding by frequency encoding (18/19)

Record a transverse (X-Z), sagittal (Y-Z), and coronal (X-Y) 2D MR image of the structure sample. Use the parameter settings of 1.

Figs. 23 a-c show the transverse (a: X-Z), sagittal (b: Y-Z), and coronal (c: X-Y) 2D MR image of the structure sample that consists of a plastic structure in oil. The settings of task 1 were used for all of the scans. Please note once again that the recorded signals are signals that were accumulated over the entire sample space and they are not signals of a thin slice. In those places where the 2D MR image is very dark, oil is hardly present in any of the locations in the sample space. If the sample space is divided into several thin slices, all of these slices include the plastic material in the dark areas of the MR image. In those places where the 2D MR image is very bright, oil is present in nearly all of the locations in the sample space. If the sample space is divided into several thin slices, all of these slices include oil in the bright areas of the MR image. In the medium-bright/medium-dark areas of the 2D MR image, a separation into several thin slices would reveal that some slice include the plastic material and others the oil. When all of the 2D MR images (transverse, sagittal, and coronal) are combined, it becomes clear that the structure sample is a sort of ladder-shaped plastic object in oil.

A: 2D spatial encoding by frequency encoding (19/19)

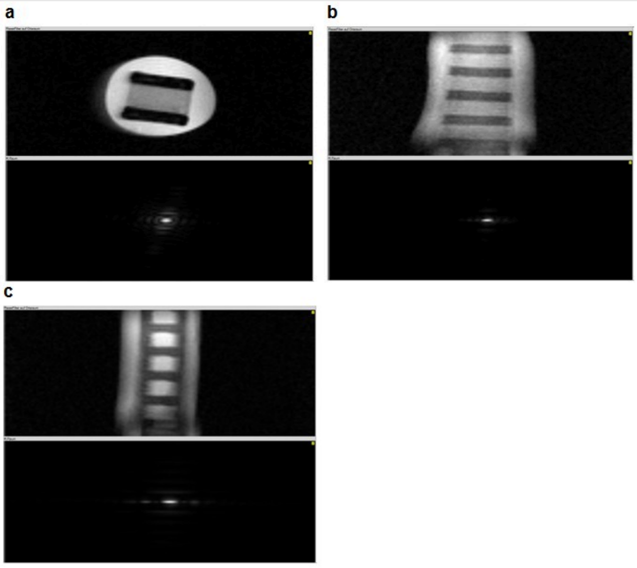

Fig. 23: Transverse (X-Z) (a), sagittal (Y-Z) (b), and coronal (X-Y) (c) MR image of the structure sample with the T_1 contrast. The images have been recorded by way of a spin echo signal (Spin Echo 2D). For this purpose, the repetition time was set to 300 ms and the echo time to 15 ms. 5 individual images were used for averaging. When all of the 2D MR images (transverse, sagittal, and coronal) are combined, it becomes clear that the structure sample is a sort of ladder-shaped plastic object in oil.

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B: Application of frequency and phase encoding (1/11)

Record a transverse 2D MR image of a double sample of oil and water by way of the rapid gradient echo method. In a transverse 2D MR image, the cross-sectional plane is parallel to the bottom of the magnetic resonance tomography scanner.

Fig. 24 shows a transverse MR image of the double sample of oil and water that is aligned in parallel to the rear edge of the magnet housing. The image has been recorded by way of a spatially encoded gradient echo signal (Flash 2D). The following settings were used for the reconstruction: repetition time: 40 ms; echo time: 5 ms; number of averages: 10; angle: 30°, readout gradient: 70.2 μTs/m, phase gradient: 70.2 μTs/m (variable in consecutive measurement sequences of an averaging step); number of image points per echo (Data points): 128; number of spatially encoded echoes (Phase steps): 64; plane of the 2D slice: X-Z slice (Slice orientation). In accordance with the theory, the recorded MR image provides a T_1 contrast, since the repetition time T_R and the echo time T_E are comparatively short. In this case, oil appears bright and water dark, i.e. the two substances of the double sample can be clearly identified.

B: Application of frequency and phase encoding (2/11)

In Fig. 24, the sample on the left is the water sample and the sample on the right is the oil sample. Please note, however, that the recorded image is a summation of the signals over the entire sample space, since an explicit slice selection with a well-defined thickness was not specified. Directly under the transverse MR image of the double sample in the position space in Fig. 24, the image of this sample in the k-space or wave-vector space is shown. The data points in the centre of the k-space determine the signal-to-noise ratio, structure, and contrast of the reconstructed image. The outer data points provide information concerning borders, edges, contours, and fine transitions.

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B: Application of frequency and phase encoding (3/11)

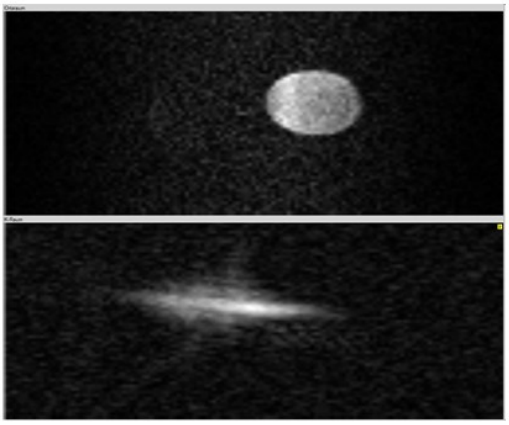

Fig. 24: Transverse (X-Z) MR image of the double sample of oil and water with the T_1 contrast. The double sample is aligned in parallel to the rear edge of the magnet housing. The image has been recorded by way of a gradient echo signal (Flash 2D). The repetition time was set to 40 ms, the echo time to 5 ms, and the excitation angle to 30°. 10 individual images were used for averaging. The two individual samples can be clearly identified in the position space (top), since oil appears bright in the T_1 contrast and water appears dark. In the wave-vector space (k-space), the data points in the centre of the k-space determine the signal-to-noise ratio, structure, and contrast in the reconstructed image. The outer data points provide information concerning borders, edges, contours, and fine transitions. The conversions between the position space and k-space are realised by way of Fourier transformations.

B: Application of frequency and phase encoding (4/11)

Record various transverse 2D MR images of the double sample with different repetition and echo times.

It is theoretically possible to produce different types of contrast with the Flash 2D method by way of a variation of the echo time and repetition time. In the theory part, we have described that the optional T_2^* contrast requires a comparatively long echo time (>18 ms). This setting is only possible with magnets with a high level of homogeneity, since the gradient echo must be generated during the FID signal and the duration of the FID depends on the homogeneity of the magnetic field. With the homogeneity of the magnet that is used here, the adjustment of a T_2^* contrast is not possible. Furthermore, we have seen that the proton-density contrast (PD) requires a comparatively long repetition time (approx. 500 ms). In principle, such a setting would also be possible for the flash method that is described here. However, due to the comparatively long repetition times, the measuring time would also become longer and the essential advantages of the flash method compared to the spin echo method would be lost. This is why the repetition time was limited to 200 ms for the Flash 2D method.

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B: Application of frequency and phase encoding (5/11)

This means that with the Flash 2D parameter settings, our compact magnetic resonance tomography scanner can only produce the T_1 contrast. This contrast can be achieved, for example, with a short echo time of 5 ms, a repetition time of 40 ms (see task 1), and an excitation angle of 30°
In principle, short echo times are always desirable, since the restored FID signal is comparatively strong in the case of short echo times. However, due to system consideration and based on special settings, the echo time is often limited in view of small values. In this system, only echo times up to 1 ms are possible.
In addition, the repetition time can be further reduced in order to produce the T_1 contrast, provided that the excitation angle is also reduced. Although the signal becomes weaker in the case of smaller excitation angles, the disturbance of the state of equilibrium is also smaller so that consecutive scans can be performed much more quickly.

B: Application of frequency and phase encoding (6/11)

Study the influence of the deflection angle on the transverse 2D MR image of the double sample of oil and water.

Figs. 25 a and b show the transverse MR image of the double sample of oil and water for an excitation angle of 10°. The following additional settings were used for the reconstruction: repetition time: 40 ms (a), 20 ms (b); echo time: 5 ms; number of averages: 10; readout gradient: 70.2 μTs/m, phase gradient: 70.2 μTs/m (variable in consecutive measurement sequences of an averaging step); number of image points per echo (Data points): 128; number of spatially encoded echoes (Phase steps): 64; plane of the 2D slice: X-Z slice (Slice orientation).

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B: Application of frequency and phase encoding (7/11)

In Fig. 25 a, the same repetition time as in task 1 was used. Based on the smaller excitation angle, the MR image in this picture is weaker than in Fig. 24 although it has been averaged. Initially, this is a disadvantage. Smaller excitation angles, however, also imply the possibility of a shorter repetition time, since the disturbance of the state of equilibrium is smaller and a large part of the longitudinal magnetisation is continuously present. Shorter repetition times, in turn, lead to a considerable reduction of the measurement duration. In Fig. 25 b, the repetition time was reduced to 20 ms. Due to the small excitation angle, the double sample can still be discerned (albeit in a rather blurry manner). The effective measuring time of the MR flash image in Fig. 25 b (12.8 s) is only half as long as in Fig. 25 a (25.6 s).

B: Application of frequency and phase encoding (8/11)

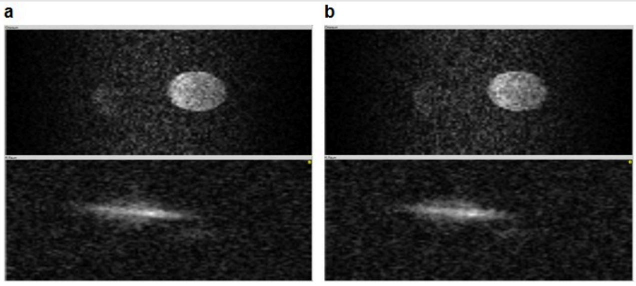

Fig. 25: Transverse (X-Z) MR images of the double sample of oil and water with the T_1 contrast. The double sample is aligned in parallel to the rear edge of the magnet housing. The images have been recorded by way of a gradient echo signal (Flash 2D). 10 individual images were used for averaging. The echo time was set to 5 ms and the excitation angle to 10°. In a, the image was recorded with a repetition time of 40 ms, while it was recorded with a repetition time of only 20 ms in b. Such a short repetition time can only provide a sufficiently strong signal if the excitation angle is comparatively small. The aim of the reduction of the repetition time is to reduce the effective measuring time. The generation of the MR image in b required only half the time as the image generation in a.

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B: Application of frequency and phase encoding (9/11)

Record a transverse (X-Z), sagittal (Y-Z), and coronal (X-Y) 2D MR image of the structure sample with acceptable quality. Try to produce these images as quickly as possible.

Figs. 26 a-c show the transverse (a: X-Z), sagittal (b: Y-Z), and coronal (c: X-Y) 2D MR image of the structure sample that consists of a plastic structure in oil. All of the images were recorded by way of a spatially encoded gradient echo signal (Flash 2D). The following settings were used for the reconstruction: repetition time: 30 ms; echo time: 3 ms; number of averages: 30; angle: 20°, readout gradient: 70.2 μTs/m, phase gradient: 70.2 μTs/m (variable in consecutive measurement sequences of an averaging step); number of image points per echo (Data points): 128; number of spatially encoded echoes (Phase steps): 64; plane of the 2D slice: X-Z slice (Slice orientation).
It can be clearly recognised that the quality is much lower compared to the 2D MR image generation by way of the spin echo method (see Fig. 23). Apparently, the influence of magnetic susceptibilities and of the inhomogeneities of the external static magnetic field $\vec{B_0}$ is considerably stronger in the case of the flash method.

B: Application of frequency and phase encoding (10/11)

Still, the Flash 2D method can be used to draw relatively good conclusions concerning the structure of the plastic sample. In addition, the measuring time is much shorter than in the case of the Spin Echo 2D method. In medical diagnostics, flash methods have become a probate mean for gaining relatively quick first impressions of the inside of the object under examination. Since, in the case of whole-body scans, the field strength is usually higher and the magnetic field homogeneity is improved, the flash method supplies rather good MR images that can be used for special diagnoses.

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B: Application of frequency and phase encoding (11/11)

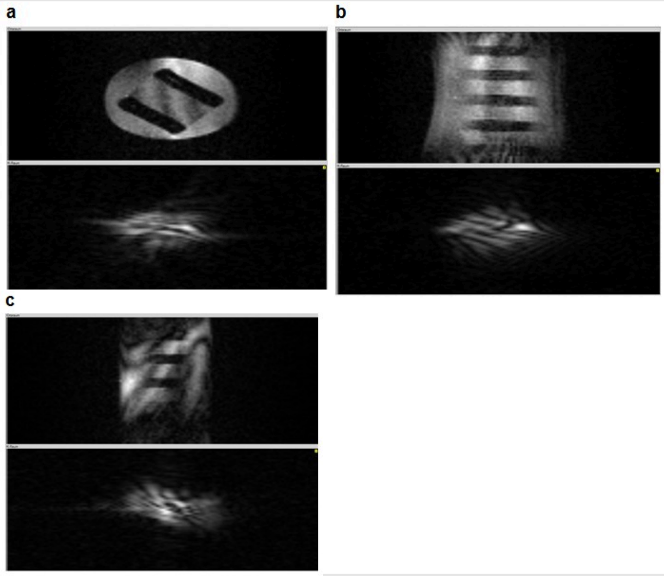

Fig. 26: Transverse (X-Z) (a), sagittal (Y-Z) (b), and coronal (X-Y) (c) MR image of the structure sample with the T_1 contrast. The images have been recorded by way of a gradient echo signal (Flash 2D). For this purpose, the repetition time was set to 30 ms, the echo time to 3 ms, and the excitation angle to 20°. 30 individual images were used for averaging. When all of the 2D MR images (transverse, sagittal, and coronal) are combined, it becomes clear that the structure sample is a sort of ladder-shaped plastic object in oil.

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