Magnetic Resonance Imaging (MRI) II

Physics | Modern Physics | Quantum physics

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

This content can also be found online at:

http://localhost:1337/c/5fea131fc68e73000314c501

General information

Application

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 (1/6)

Prior knowledge

The prior knowlege required for this experiment is found in the theory section or in P5942500.

Main principle

The experiment group "Magnetic resonance imaging II" is an extension of the experiment group "Magnetic resonance imaging I". This means that the fundamental principles for the comprehension of the following experiments have already been described and explained in detail. The aim of these experiments is to show how the spin echo technique can be used to generate 2D MR images of a slice of a well-defined thickness, orientation, and size (Localized Spin Echo 2D). These parameters determine the so-called "field of view" (FOV) of the MR image. We will introduce a method that enables the recording of 3D MR images (Spin Echo 3D). For this purpose, an additional phase encoding will be performed in the third dimension. Both methods include the automatic calibration of the system frequency with regard to the Larmor frequency. As a result, the MR image is more stable over several averaging steps

Other information (2/6)

Tasks

A: Generation of 2D MR images with explicit slice selection (Localized Spin Echo 2D)

1. Record a transverse (X-Z) 2D MR image of the structure sample in oil with a well-defined slice thickness (e.g. 10 mm). In a transverse 2D MR image, the cross-sectional plane is parallel to the bottom of the magnetic resonance tomography scanner.
2. Generate transverse (X-Z) 2D MR images of the structure sample in oil with different slice thicknesses. Please note that the thinner the slice is, the weaker the signal of the generated images will be. This means that, for an adequate MR image of a thin slice, the number of averages must be higher than in the case of thicker slices.
3. Study the effect of the number of averages on the 2D MR image of a very thin transverse (X-Z) slice of the structure sample (1 mm).

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Other information (3/6)

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Tasks

4. Generate a sagittal 2D MR image of the structure sample in oil in which the plastic ladder appears black. Apart from a specific position of the structure sample in the sample chamber, a specific orientation of the "field of view" (FOV) must also be selected.
5. Then, generate the corresponding coronal 2D MR image while maintaining the alignment of the structure sample in the sample chamber (see point 4). Why does the ladder now appear bright? Answer this question by having a closer look at the structure sample.
6. Following a 45° rotation of the structure sample in the sample chamber, try to generate an analogous 2D MR image like in task 1 by way of the setting of the orientation of the "field of view".

Other information (4/6)

PHYWE

Tasks

7. Record 2D MR images of all kinds of samples with thin slice thicknesses. Beans, buds, and branches are particularly suitable. It is also possible to produce MR images of small insects with good resolution. Please note that, in the case of biological samples, the repetition time must be much longer than in the case of the structure sample. In addition, a very large number of averages is required for an adequate sectional MR image. This is why the total effective measuring time is clearly longer.

B: Generation of 3D MR images (Spin Echo 3D)

1. Record a 3D MR image of the structure sample with a comparatively small number of data points and phase steps (e.g. 16 in all of the three encoding directions).
2. Record a 3D MR image of the structure sample with a comparatively large number of data points and phase steps (e.g. 64 in all of the three encoding directions).

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Other information (5/6)

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Tasks

3. Examine the influence of the number of averages on the 3D MR image of the structure sample that was generated in 1.
4. Record 3D MR images of all kinds of samples. Beans, buds, and branches are particularly suitable. Please note that, in the case of biological samples, the repetition time must be much longer than in the case of the structure sample. In addition, a larger number of averages is useful for obtaining an adequate MR image. This is why the total effective measuring time is clearly longer.
5. Think about how the total measuring time for the generation of a 3D MR image is calculated. Take into consideration that the measuring time is mainly determined by the two phase encoding processes.

Other information (6/6)

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Note:
The experiments A and B of the experiment ensemble "Magnetic resonance imaging II" are no longer part of the basic course within the "measure MRT" software. In order to perform the experiments, the course "Imaging 2" must be selected. This course includes 3 lessons. The "Localizer" lesson enables the rapid generation of a 2D MR image with parameters that are preset to a large extent. It can be used to check whether a sample can actually provide an MR image, i.e. whether the sample is actually penetrated by the external magnetic field (this can be useful when using own samples). The experiments A and B are performed under the lessons "Localized Spin Echo 2D" and "Spin Echo 3D".

Note: In order to select the course "Imaging 2", click the plus sign in the parameters area.

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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 (1/21)

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The preceding experiment ensembles have shown that the nuclear spins align in a static magnetic field $\vec{B}_0$, The slightly preferred parallel alignment (compared to the antiparallel alignment) leads to an effective longitudinal magnetisation $\overrightarrow{M_{L0}}$ 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 (2/21)

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 (3/21)

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). As a matter of fact, the FID signal decays in a time T_2^* ithat is considerably shorter than the relaxation time $T_2 (T_2^* < T_2 < T_1)$. This is due to purely static field inhomogeneities that cause the artificial, yet systematic, fanning-out of the nuclear spin ensemble. The artificial dephasing can be partly undone by way of a 180° pulse after the time T_S. The signal, which is thereby restored after the echo time $T_E = 2 \cdot T_S$ is called the spin echo. 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. Particularly evident is the selection of a specific slice by way of the so-called magnetic slice selection gradient (e.g. generated by 2 opposite pairs of coils). Since, due to the slice selection gradient, the static magnetic field is increased at one gradient coil and decreased at the opposite coil, the individual nuclear spins precess with different speeds at different locations, i.e. they show resonance at different frequencies.

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Theory (4/21)

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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 $\Delta\omega_0$ of the HF pulse in combination with the strength of the gradient field (see Fig. 1). Since the bandwidth $\Delta\omega_0$ of an HF pulse is proportional to the inverse pulse duration, it is ultimately the pulse duration and the gradient strength that determine the thickness of the selected slice. 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. This was the case in all of the preceding experiments concerning magnetic resonance imaging. In the following experiment ensemble "Magnetic resonance imaging II", the slice selection gradient is explicitly used for the selection of slices. However, it is slightly modified. In the gradient, the magnetisation, which is excited during the slice selection pulse, dephases. This leads to signal losses. In order to counteract these signal losses, an additional gradient with the opposite sign is applied at the end of the selection pulse. If the integral of this gradient corresponds to half of the slice selection gradient, the unwanted dephasing of the nuclear spin ensemble in the excited slice can be avoided. As a result, the signal from the excited slice remains comparably strong.

Theory (5/21)

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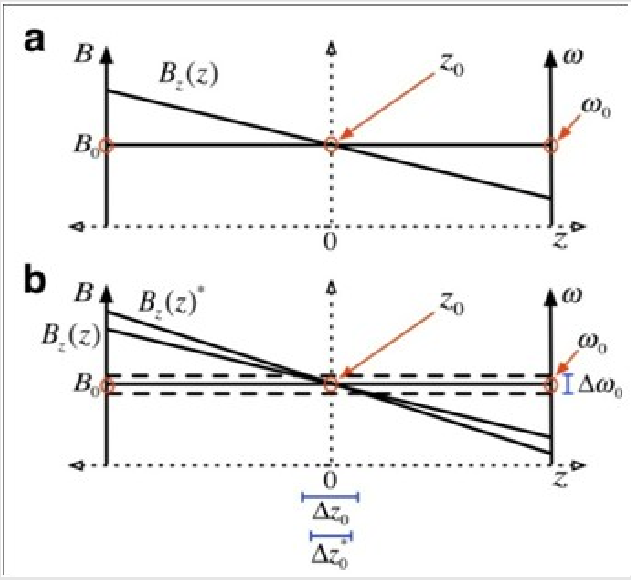

Fig. 1: Gradient field for the slice selection in the z-direction. Figure (a) shows that, due to the position-dependent gradient or due to the new, position-dependent magnetic field B_z (z), the nuclear spins come into resonance with the HF pulse with the frequency ω_0 only at the point z_0. Figure (b) 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.

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Theory (6/21)

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In addition, we have also learnt from the experiment ensembles "Spatial encoding in nuclear magnetic resonance" and "Magnetic resonance imaging I" that signals are assigned to their exact location of origin by way of magnetic gradient fields. Gradient fields also enable the spatial encoding of a slice and, thereby, the actual generation of the MR image. This is where the concept of voxels and resolution comes into play. The voxels (3D) correspond to the individual discrete image elements of an MR image (just like the pixels in the case of 2D). 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. In order to understand this assignment, we try at first to enforce 1D spatial encoding, i.e. the encoding of a voxel strip via a spin echo signal (see Fig. 2).

Theory (7/21)

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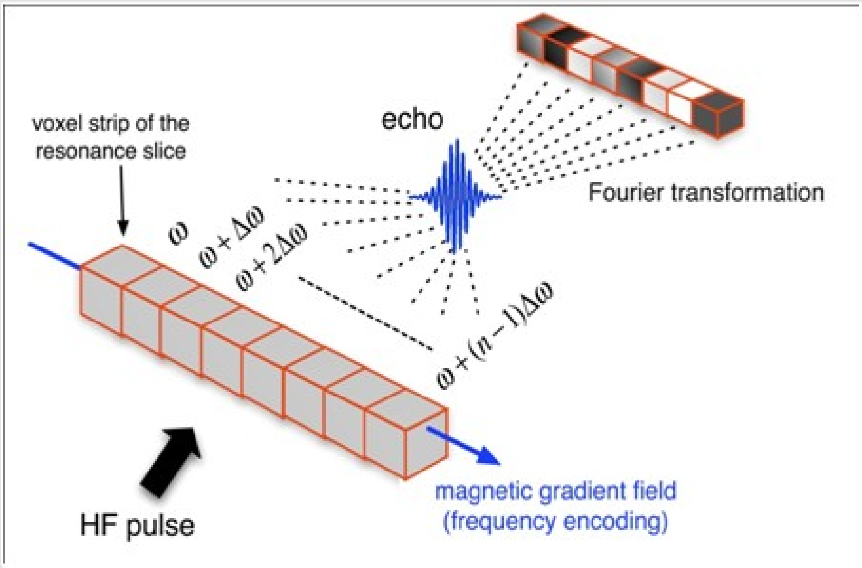

Fig. 2: Spatial encoding of a 1D voxel strip (n voxels). During the measurement of the spin echo, a linear magnetic field gradient is applied in the encoding direction. This gradient causes the nuclear spins of the voxel strip to precess with a frequency that increases linearly (frequency encoding). 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.

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Theory (8/21)

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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.

Theory (9/21)

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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.

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Theory (10/21)

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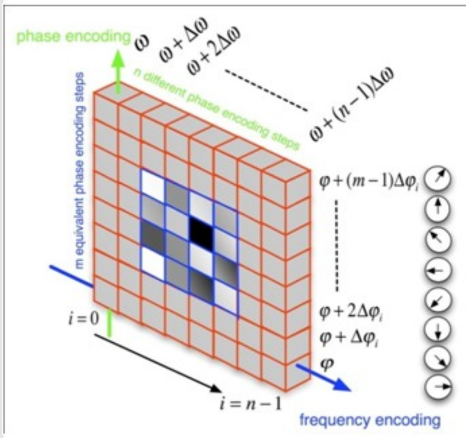

Fig. 3: Spatial encoding of a 2D voxel slice (n x m voxels). During the measurement of the spin echo, a linear magnetic field gradient is applied in one of the two encoding directions (e.g. in the x-direction). This gradient causes the nuclear spins along this direction to precess with a fre-quency that 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.

Theory (11/21)

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Let us look at the example of a matrix of n x m voxels. In one direction, we obtain simple spatial encoding by way of a frequency encoding gradient. In the other direction, we need m spin echoes with different phase encoding, i.e. m phase encoding steps, for the spatial encoding of m voxels. The m 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 n x m 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).

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Theory (12/21)

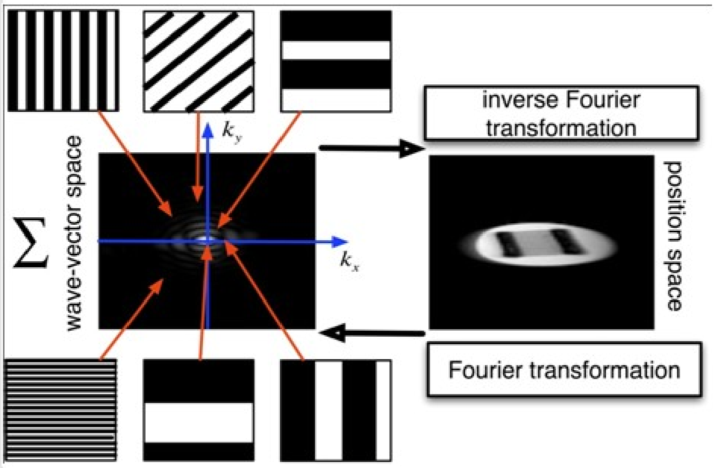

Fig. 4: 2D imaging in the k-space and position space. In the k-space, every point corre-sponds to a stripe pattern with a well-defined frequency. The raw data values in the k-space determine the weighting of these stripe patterns and, thereby, the wave-vector image. Coarser strip patterns are very close 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.

Theory (13/21)

Based on our preliminary considerations, the sequence for the 2D MR imaging of a slice can now be easily comprehended (see Fig. 15).
The starting point for our considerations is the state of equilibrium, i.e. the preferred alignment of the nuclear spins parallel to the static magnetic field $\vec{B}_0$. A 90° HF pulse that fulfils the resonance condition (1) deflects the excess spins by 90° into the plane perpendicular to the static magnetic field $\vec{B}_0$
If this pulse is superimposed by the slice selection gradient (e.g. in the z-direction), only the spin in a certain slice will be deflected. The thickness of this slice depends on the bandwidth of the pulse (inverse pulse duration) and on the strength of the slice selection gradient. Finally, the envelope of the HF pulse (excitation profile) determines the precise nature of the selected slice. For example, in order to select a cuboid as a slice, a sinc pulse is used, since, in accordance with the rules of the Fourier transformation, the frequency spectrum of a sinc pulse is a square.

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Theory (14/21)

If only the slice selection gradient was applied during the 90° pulse, an unwanted dephasing of the magnetisation in the gradient would result, i.e. the signal would be considerably weaker. In order to counteract these signal losses, an additional gradient with the opposite sign is applied at the end of the selection pulse. If the integral of this gradient corresponds to half of the slice selection gradient, the unwanted dephasing of the nuclear spin ensemble in the excited slice can be avoided. As a result, the signal from the excited slice remains comparably strong.
The magnetisation vector that results from the deflected spins in the selected slice, continues to precess around $\overrightarrow{B_0}$ until the nuclear spin ensemble dephases during the time $T_2^* < T_2$. The result is the FID signal. Since the transverse magnetisation, which evokes the FID signal, does not relax until after the time T_2 and since the dephasing 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 the spin echo). The 180° pulse no longer needs to be superimposed by the slice selection gradient, since, even after the global 180° inversion, the undeflected nuclear spins do not contribute to the signal. This means that the 180° inversion of the nuclear spins is only important for the nuclear spins that have already been deflected.

Theory (15/21)

The spin echo signal after the time $T_E = 2 \cdot T_S$ is now superimposed by the frequency encoding gradient (readout gradient). It evokes spatial encoding in the first encoding direction of the selected slice (e.g. in the x-direction). In addition, a gradient thus applied also causes an unwanted dephasing of the nuclear spin ensemble, which leads to a decrease in signal strength of the spin echo that is to be measured. However, this unwanted side effect can be counteracted in a particularly easy way. If, prior to the 180° pulse, a magnetic gradient of half the duration τ_{FG} of the final readout gradient $(2 \cdot \tau_{FG})$ is applied in the first encoding direction (dephasing gradient), the nuclear spin ensemble will dephase before the spin echo echo measurement signal and into the direction that is opposite to 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. Simultaneous to the dephasing gradient, the phase encoding gradient is applied in the second encoding direction (e.g. in the y-direction). Its amplitude depends on the measurement cycle, i.e. the resolution of m voxels in the phase encoding direction requires m different amplitudes (m different phase encoding gradient) in m successive measurement cycles. The spin echo signals of the m measurement cycles then 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 (16/21)

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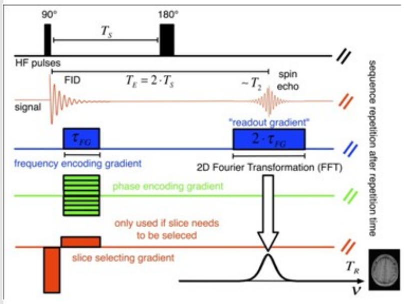

Fig. 5: Recording of a 2D MR image by way of a spin echo (Spin Echo 2D). The slice selection gradient is applied during the 90° HF pulse. This gradient brings only the nuclear spins of a certain slice with a corresponding slice thickness into resonance (red). During the spin echo, the frequency encoding 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 ($2\tau_{FG}$) is applied prior to the 180° pulse. Spatial encoding of the second direction of the selected resonance slice is ensured by the phase encoding gradient (green). The resolution in the phase encoding direction determines the measuring time to a very large extent, since the sequence that is shown must be repeated with different phase encodings in accordance with this resolution.

Theory (17/21)

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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_E between consecutive measurements with different phase encoding gradients. These two parameters are used to produce different contrasts of the MR image.
The different contrast methods have been presented in the experiment ensemble "Magnetic resonance imaging I". The parameter settings for a certain contrast, which were found during these experiments, are mainly based on the discernibility of pure oil and pure water. Oil has a very short relaxation time. As a result, a rather short repetition time can also be selected, e.g. in the T_1 contrast, in order to evoke good discernibility, e.g. with regard to water. The relaxation time of the chemical constituents of biological samples is usually considerably longer than the relaxation time of pure oil. This is why, for the generation of MR images of biological samples, the repetition times must be set to considerably higher values in order to produce good MR contrast.

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Theory (18/21)

The sequence of Figure 5 can be used to record the 2D MR image of any slice and to reconstruct it via a 2D Fourier transformation. If several 2D slices are recorded that are superimposed on one another, the individual slices can also be used to generate a 3D MR image. 3D images offer important advantages compared to 2D images. Information concerning the exact volume of organs, for example, can evoke a better understanding concerning the way they work. In addition, the 3D representation also enables the quick identification of artefacts and a clearer separation of the individual constituents of the MR sample. This is an enormous advantage in medical diagnosis, e.g. for the preparation of surgeries or the exact localisation of tumours. A disadvantage of the 3D representation is the considerably longer measuring time, of course.
3D MR images can also be generated in one single step and not by the arduous composition of 2D MR images of several individual slices. In this context, we will once again look at the sequence for the generation of 2D MR images of a single slice (see Fig. 5.). If we perform additional phase encoding in the third dimension, in place of the slice selection gradient, and if this phase encoding is applied simultaneously to the phase encoding in the second dimensions, the final spin echo signal will actually be spatially encoded in all three spatial directions.

Theory (19/21)

This means that 3D MR images require one frequency encoding and two phase encodings (see Fig. 6).
Depending on the resolution, several phase encoding steps are required in the two phase encoding directions. Let us look at the example of a cube of 4 x 4 x 4 voxels. One slice of the cube requires 4 different phase encoding steps of the regular phase encoding gradient. For the recording of all 4 slices, 4 additional phase encoding steps are required for the additional phase encoding gradient.
If the resolution is doubled in one of the two phase encoding directions, the measuring time will also be doubled (see "Spatial encoding in nuclear magnetic resonance"). However, if the resolution is doubled in the frequency encoding direction, the measuring time will not be doubled (see "Spatial encoding in nuclear magnetic resonance"). The assignment of the corresponding grey values to a cube of 8 x 8 x 8 voxels requires 4 times the measuring time as a cube of 4 x 4 x 4 voxels.

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Theory (20/21)

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The actual reconstruction of the 3D MR image is realised by way of a three-dimensional Fourier transformation. It should be noted that the sequences for 2D and 3D MR imaging are very similar. Consequently, the parameter settings for a certain contrast must be selected in complete analogy (repetition time, echo time). However, due to the additional phase encoding, 3D MR imaging requires a longer measuring time.

Theory (21/21)

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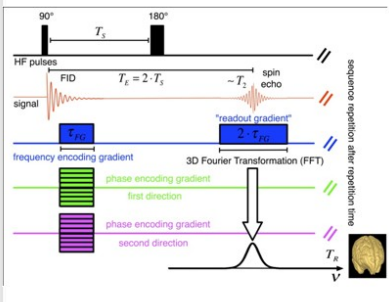

Fig. 6: Recording of a 3D MR image by way of a spin echo (Spin Echo 3D). During the spin echo, the frequency encoding gradient (blue) is applied. It evokes spatial encoding in one of the three directions of the 3D space ("readout gradient"). In order to counteract a dephasing of the spins at the readout moment, an additional "dephasing gradient" that is half as long (τ_{FG}) as the readout gradient (\(2\tau_{FG\}) is applied prior to the 180° pulse. The second and third directions of the 3D space are spatially encoded by way of phase encoding gradients (green, pink). The resolutions in the two phase encoding directions determine the measuring time to a very large extent, since the sequence that is shown must be repeated with different phase encodings in accordance with this resolution.

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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. 7. 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. 9). 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. 9).

Fig. 7: Set-up of the MRT training unit

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Setup (2/2)

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Connect the control unit and the magnet by way of the gradient and BNC cables that are intended for this purpose (see Fig. 8). Then, connect the USB interfaces of the control unit and measurement computer via a USB 2.0 high-speed cable (see Fig. 9). 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. 8: Magnet and control unit connectors

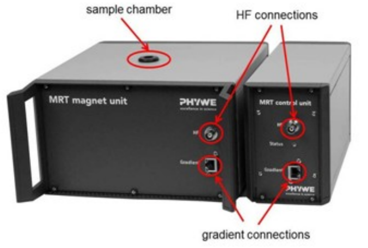

Fig. 9: Connectors at the back of the control unit

Procedure (1/14)

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When the "measure MRT" software is started, a window will open automatically as shown in Fig. 10. 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".

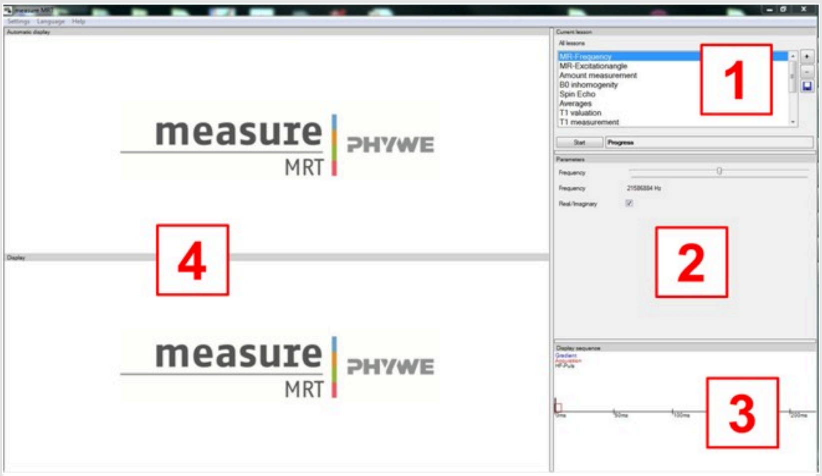

Fig. 10: Areas of the "measure MRT" program

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Procedure (2/14)

Note: The following experiments (A and B) should be performed in chronological order. In order to be able to do this, the experiment ensembles "Fundamental principles of nuclear magnetic resonance" (P5942100), "Relaxation times in nuclear magnetic resonance" (P5942200), "Spatial encoding in nuclear magnetic resonance" (P5942300), and "Magnetic resonance imaging I" (P5942400) should have been performed. If this is the case, the corresponding settings have been saved. A special feature of the experiment ensemble "Magnetic resonance imaging II" (P5942500) is the automatic calibration of the system frequency. The system frequency will be checked prior to every measurement. If necessary, it will be corrected, i.e. it will be adapted once again to the Larmor frequency of the hydrogen protons. The advantage of this is that the imaging process over longer period of times provides a high-quality result. Measurements with a long duration are often subject to temperature fluctuations. Even small temperature changes affect the magnetic field and, thereby, also the Larmor frequency. If the system frequency is not corrected accordingly, this will lead to the "smearing" of the recorded image in the direction of the frequency encoding gradient (readout gradient). The automatic system frequency calibration enables a trouble-free image generation over several averaging steps. Like in the case of all of the preceding experiment ensembles, it is often useful to check the homogeneity of the magnetic field and to improve it by way of the so-called magnetic field shims.

Procedure (3/14)

A: Generation of 2D MR images with explicit slice selection (Localized Spin Echo 2D)

1. Place the 10 mm structure sample into the sample chamber of the unit. Align the sample so that the openings of the ladder are oriented towards the back of the magnet. In the experiments area (lessons), select the lesson Localized Spin Echo 2D. You can find this lesson in the course "Imaging II" (see the note). In the parameters area, the tab "Encoding Contrast Field of View (FOV)" now shows the setting options Data points, Phase steps, Averages, Repetition time, Echo time, FOV read, FOV phase, and Slice thickness (see Fig. 11). The button "Select orientation" opens a window in which the position, size, and orientation of the scanning window can be adjusted more precisely. It offers the setting options Alpha, Beta, Gamma, Field of View (FOV), FOV phase, FOV read, Slice thickness, Offset phase, Offset read, and Offset slice (see Fig. 12). The scanning window that is adjusted via the parameters is displayed in graphical form in a Cartesian system of coordinates. The tab "Advanced" offers additional setting options for advanced users. Here, the Sample frequency and the spoilers (spoiler X, spoiler Y, spoiler Z) can be adjusted for all of the gradients (see Fig. 13). The default values can be maintained for all of the experiments.

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Procedure (4/14)

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To begin with, set the Data points (number of image points per echo) and Phase steps (number of spatially encoded echoes) to 64. Set the Repetition time to 150 ms, the Echo time to 10 ms, and the number of Averages to 1. Then, define the "Field of View". Select an FOV read of 20 mm, an FOV phase of 20 mm, and a Slice thickness of 10 mm. The recording of a purely transverse (X-Z) slice can be specified by way of the button Select orientation. Set Alpha and Gamma to 90° and Beta to 0°. Furthermore, the scanning window should be aligned centrally, i.e. without any shift in one of the three spatial directions (default = 0 mm). Click the Save button.
Start the measurement.

2. Maintain the settings of 1. Vary the slice thickness via the setting option Slice thickness. In the case of thinner slices, increase the number of averaging steps via the Averages slider. Record the 2D MR image of a 5 mm slice and of a 2 mm slice with 5 and 20 averages, respectively (attention: longer measuring time required).

Procedure (5/14)

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3. Maintain the settings of 1 and explicitly study the influence of the number of averages on the 2D MR image of transverse 1 mm slice of the structure sample. Generate the 2D MR images of the 1 mm slice with 1, 5, and 50 averages (attention: longer measuring time required).

4. Use the settings of 1 again. Now, set the number of averages to 10 by way of the Averages slider. Check the alignment of the structure sample in the sample chamber of the magnet once again. The sample should be aligned so that the openings of the ladder are oriented towards the back of the magnet. Click the button "Select orientation" and set the angle for Alpha to 0° and the angles for Beta and Gamma to 90°. With these values, the scanning window is aligned in parallel with regard to the back of the magnet housing. Confirm this window with Save. Start the measurement. The ladder now appears black.

5. Maintain the settings of 4. Click the button "Select orientation" and set the angle for Beta to 0°. With these values, the scanning window is aligned perpendicular to the back of the magnet housing. Confirm this window with Save. Start the measurement. The ladder now appears bright.

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Procedure (6/14)

6. Use the settings of 1 again (Alpha = 90°, Beta = 0°, Gamma = 90°). Set the number of averages to 1. Rotate the structure sample in the sample chamber by 45° and record a transverse (X-Z) 2D MR image. Click the button "Select orientation" and adjust the angle for Gamma so that the transverse 2D MR image of the structure sample that will be generated has the same orientation as in 1 (note: there will be several results).
7. Insert a biological sample into the empty test tube (10 mm). Beans, buds, and branches are particularly suitable. When inserting the test tube into the sample chamber of the magnet, ensure that the object to be examined is positioned in such a way that it is actually penetrated by the external magnetic field. The volume of the test tube, which is penetrated by a very homogeneous magnetic field after the insertion into the sample chamber, is marked with 2 lines. The sample to be examined is ideally located centrally between the upper and lower mark. In order to position the object optimally in the sample chamber, it may be useful to cut a thin branch or twig (or alternatively a piece of plastic) to size and to use it a support for the object. It is, of course, relatively easy to generate sectional images of longer samples, e.g. thin branches or twigs, whose length corresponds approximately to that of the test tube.

Procedure (7/14)

In this case, the magnetic field will definitely penetrate a certain volume of the sample in the sample chamber.
Insert the test tube with the correctly positioned biological sample into the sample chamber of the magnet. To begin with, set the Data points (number of image points per echo) and Phase steps (number of spatially encoded echoes) to 128. Then, set the number of Averages to a relative high value, e.g. 500 (guide value). Select also a relatively long Repetition time, e.g. 750 ms (guide value). Set the Echo time to 12 ms. Then, define the "Field of View" (FOV). Set the FOV read and FOV phase to 12 mm and the Slice thickness to only 1 mm. You can view the scanning window under the button "Select orientation". For a transverse slice (X-Z) of the biological sample, set Alpha to 90°, Beta to 0°, and Gamma to 90°. Start the measurement (attention: a very long measuring time is required).

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Procedure (8/14)

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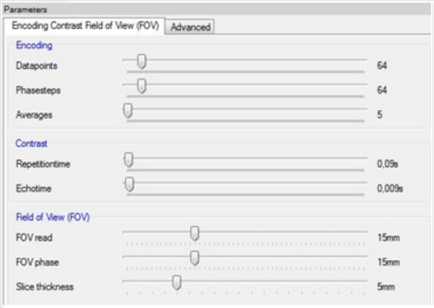


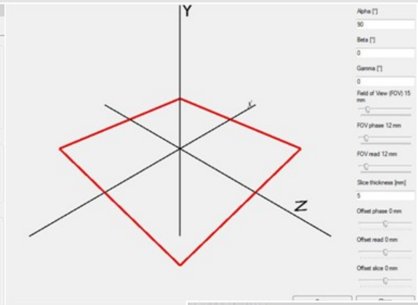
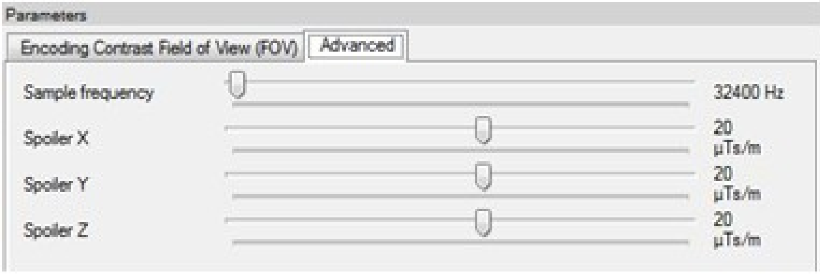


Fig. 12: Localized Spin Echo 2D – parameters (FOV)

Fig. 13: Localized Spin Echo 2D – parameters (advanced)

Fig. 11: Localized Spin Echo 2D – parameters (general)



Procedure (9/14)

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B: Generation of 3D MR images (Spin Echo 3D)

1. Place the 10 mm structure sample into the sample chamber of the unit. In the experiments area (lessons), select the lesson Spin echo 3D. You can find this lesson in the course "Imaging II" (see the note). In the parameters area, the tab "Encoding Contrast Field of View (FOV)" now shows the setting options Data points, Phase steps, Second phase steps, Averages, Repetition time, Echo time, FOV read, FOV phase, and FOV phase 2 (see Fig. 14). The tab "Advanced" offers additional setting options for advanced users. Here, the Sample frequency and the spoilers (spoiler X, spoiler Y, spoiler Z) can be adjusted for all of the gradients (see Fig. 15). The default values can be maintained for all of the experiments.
   To begin with, set the Data points (number of image points per echo), Phase steps (number of spatially encoded echoes), and Second phase steps to 16. Set the Repetition time to 150 ms, the Echo time to 10 ms, and the number of Averages to 1. Then, define the "Field of View". Select an FOV read of 15 mm, an FOV phase of 15 mm, and an FOV phase 2 of 15 mm. Start the measurement.

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Procedure (10/14)

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2. Use the settings of 1 again. Set the Data points (number of image points per echo), Phase steps (number of spatially encoded echoes), and Second phase steps to 64. Start the measurement (attention: a longer measuring time is required). Note: The reconstruction of the 3D MR image will take several minutes.
3. Use the settings of 1 again (Data points, Phase steps, and Second phase steps = 16). Now, record two 3D MR images with 3 and 10 averages, respectively (attention: longer measuring time required).
4. Insert a biological sample into the empty test tube (10 mm). Beans, buds, and branches are particularly suitable. When inserting the test tube into the sample chamber of the magnet, ensure that the object to be examined is positioned in such a way that it is actually penetrated by the external magnetic field. The volume of the test tube, which is penetrated by the magnetic field after the insertion into the sample chamber, is marked with 2 lines. The sample to be examined is ideally located centrally between the upper and lower mark. In order to position the object optimally in the sample chamber, it may be useful to cut a thin branch or twig (or alternatively a piece of plastic) to size and to use it as support for the object.

Procedure (11/14)

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It is, of course, relatively easy to generate 3D MR images of longer samples, e.g. thin branches or twigs, whose length corresponds approximately to that of the test tube. In this case, the magnetic field will definitely penetrate a certain volume of the sample in the sample chamber.
Insert the test tube with the correctly positioned biological sample into the sample chamber of the magnet. To begin with, set the Data points (number of image points per echo), Phase steps (number of spatially encoded echoes), and Second phase steps (number of spatially encoded echoes) to 32. Then, set the number of averages to 10, for example (guide value). Select a relatively long Repetition time, e.g. 1 s (guide value). Set the Echo time to 12 ms. Then, define the "Field of View" (FOV). Set the FOV read, FOV phase, and FOV phase 2 to 15 mm. Start the measurement (attention: a very long measuring time is required). Note: The reconstruction of the 3D MR image will take some time.

5. Read the calculated measuring duration with a varying number of Data points, Phase steps, and Second phase steps. The calculated measuring time will be displayed in the lower area of the parameters area. If all the other parameter settings are maintained, a rule concerning the actual duration of the measurement can be deduced rather easily.

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Procedure (12/14)

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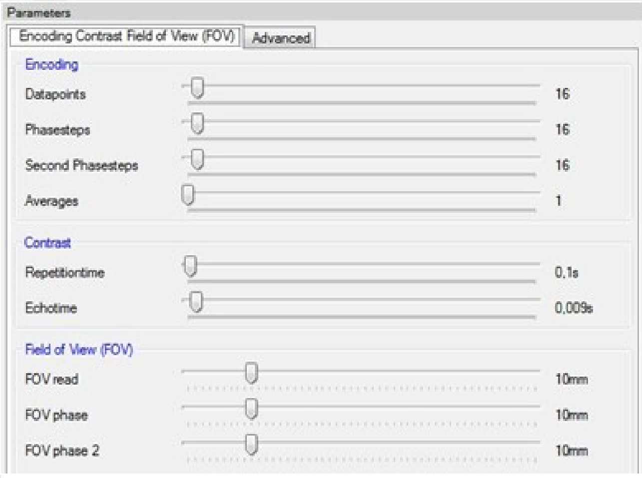

Fig. 14: Spin Echo 3D – parameters (general)

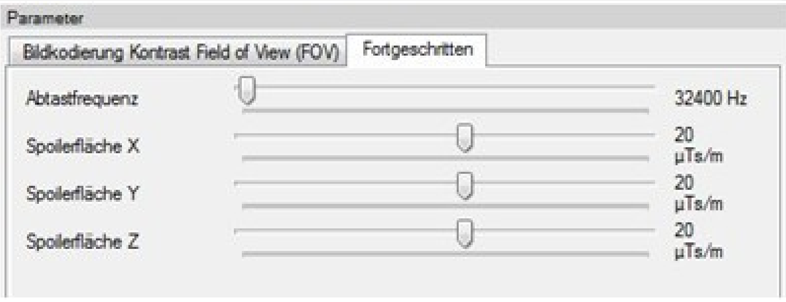

Fig. 15: Spin Echo 3D – parameters (advanced)

Procedure (13/14)

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3D results panel After the successful completion of a 3D measurement, a so-called 3D results panel will open automatically. It shows the 3D MR image in a spatially resolved manner. In some aspects, this results panel differs from the 2D representation in the standard measurements area of the "measure MRT" software and it offers additional setting options. The horizontal slider (see Fig. 16) can be used to adjust the threshold value of the noise level. The grey values that will be used in the visualisation algorithm for the representation of the 3D MR image are specified here. If the threshold value is too low, the 3D MR representation of the examined object will be lost in the noise. If, on the other hand, the threshold is too high, the number of grey values that are permitted is too low and the 3D MR image of the examined object will be incomplete. The check box in the lower right-hand corner of the 3D results panels can be used to activate or deactivate the 3D representation. If this check box is selected, the MR image will be displayed in a three-dimensional manner by way of interpolations. If it is deactivated, the recorded slices (FOV read x FOV phase) will be displayed individually. The vertical slider with the "magnifying glass" symbol adjusts the zoom. The two vertical sliders with the "arrow" symbols can be used to place a horizontal or vertical cut through the sample. The examined 3D object can be rotated with the aid of the mouse. To do so, click the object and keep the mouse button pressed.

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Procedure (14/14)

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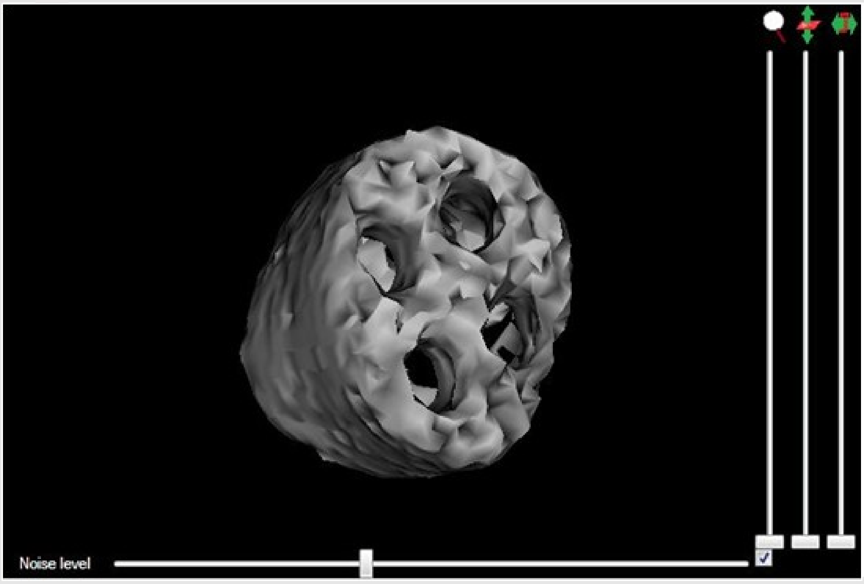

Fig. 16: 3D results panel

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Evaluation

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A: Generation of 2D MR images (1/16)

Record a transverse (X-Z) 2D MR image of the structure sample in oil with a well-defined slice thickness (e.g. 10 mm). In a transverse 2D MR image, the cross-sectional plane is parallel to the bottom of the magnetic resonance tomography scanner.

Fig. 17 shows the transverse MR image of a 10 mm slice of the structure sample in oil. The structure sample was aligned so that the openings of the ladder (structure) were oriented towards the back of the magnet. The MR image of the slice has been recorded by way of a spatially encoded spin echo signal (Localized Spin Echo 2D). The following settings were used for the imaging process: repetition time: 150 ms; echo time: 10 ms; number of averages: 1; data points: 64; phase steps: 64; FOV read: 20 mm; FOV phase: 20 mm; slice thickness: 10 mm; alpha: 90°; beta: 0°; gamma: 90°. FOV read and FOV phase have the same function as the sliders "Read gradient" and "Phase gradient" in the experiment "Spin Echo 2D" (see "Magnetic resonance imaging I"). For the "Field of View" (FOV), both values were calibrated for the coil size so that the adjustment is performed with mm as the unit. Alpha, Beta, and Gamma determine the exact alignment of the scanning window in space. The echo time T_E and the repetition time T_R determine the contrast of the MR image. Since this experiment part uses the structure sample (plastic in oil), it is possible to select a relatively short repetition time. With 150 ms, the repetition time is above the relaxation time of oil (approximately 120 ms).

A: Generation of 2D MR images (2/16)

The image in Fig. 17 corresponds to a summation of the signals over the explicitly selected slice of the sample. Directly under the transverse MR image of the structure 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.

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A: Generation of 2D MR images (3/16)

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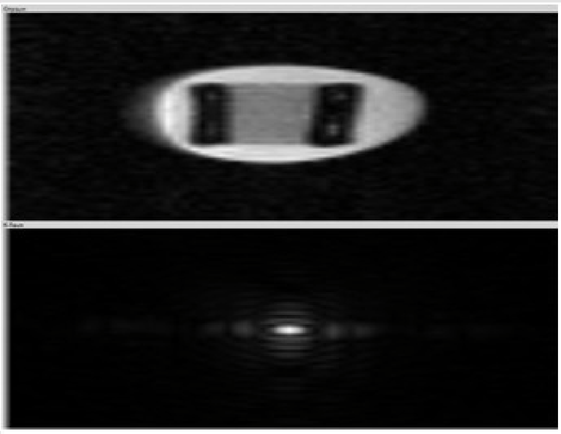

Fig. 17: Transverse (X-Z) MR image of a 10 mm slice of the structure sample in oil. The openings of the ladder are aligned so that they are oriented towards the back of the magnet. The image has been recorded by way of a spatially encoded spin echo signal (Localized Spin Echo 2D). For this purpose, the repetition time was set to 150 ms and the echo time to 10 ms. The "Field of View" determines the precise scan-ning window for the image generation of the trans-verse slice. In order to record the transverse MR image, the parameters of the "Field of View" were set to the following values: FOV read: 20 mm, FOV phase: 20 mm, slice thickness: 10 mm, alpha: 90°, beta: 0°, gamma: 90°.

A: Generation of 2D MR images (4/16)

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Generate transverse (X-Z) 2D MR images of the structure sample in oil with different slice thicknesses. Please note that the thinner the slice is, the weaker the signal of the generated images will be. This means that, for an adequate MR image of a thin slice, the number of averages must be higher than in the case of thicker slices.

Figs. 18 a and b show the transverse MR image of the structure sample in oil for two different slice thicknesses (a: 5 mm, b: 2 mm). In a, the resulting MR image has been averaged based on 5 individual measurements, while the image in b has been averaged based on 20 individual measurements. All the other settings were the same as in task 1. Both MR images (a and b) illustrate the nature of the structure sample rather well in the transverse plane. In the very bright areas, oil is present across the entire selected slice. In the black areas, the plastic material (ladder beams) is present across the entire selected slice. The areas in the middle of the MR images are more or less grey. In these areas, plastic and oil are present across the entire selected slice (ladder spokes and gaps between the spokes). It can be clearly noticed that this central area is slightly brighter in Fig. 18 b than in Fig. 18 a. This indicates that there is more oil in the central area of the selected 2 mm slice of Fig. 18 b than in the central area of the selected 5 mm slice in Fig. 18 a.

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A: Generation of 2D MR images (5/16)

In addition, both MR images (18 a and b) have a similar level of quality, i.e. both have a similar signal-to-noise ratio (SNR). This was achieved by the higher number of averages during the recording of the thinner 2 mm slice. Of course, a higher number of averages prolongs the recording time.

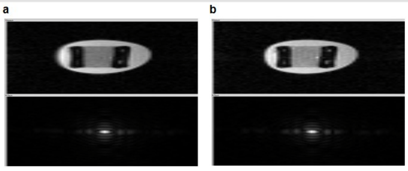

Fig. 18: Transverse (X-Z) MR image of the structure sample in oil for two different slice thicknesses (a: 5 mm, b: 2 mm) after different numbers of averaging steps (a: 5, b: 20). The openings of the ladder are aligned so that they are oriented towards the back of the magnet. The image has been recorded by way of a spatially encoded spin echo signal. For this purpose, the repetition time was set to 150 ms and the echo time to 10 ms. In the very bright areas, oil is present across the entire selected slice. In the black areas, the plastic material is present across the entire selected slice. The areas in the middle of the MR images are more or less grey. In these areas, plastic and oil are present across the entire selected slice (ladder spokes and gaps between the spokes). In Fig. 18 b, this central area is slightly brighter than in Fig. 18 a. This indicates that there is more oil in the central area of the selected 2 mm slice of Fig. 18 b than in the central area of the selected 5 mm slice in Fig. 18 a.

A: Generation of 2D MR images (6/16)

Study the effect of the number of averages on the 2D MR image of a very thin transverse (X-Z) slice of the structure sample (1 mm).

Figs. 19 a-c explicitly show the influence of the number of averages on the 2D MR image of a transverse 1 mm slice of the structure sample. The most important settings were adopted from task 1. Obviously, the quality of the MR image of a very thin slice, i.e. the signal-to-noise ratio of this slice, increases when the number of averages increases. With only one recorded image (1 average), the resulting 2D MR image of the thin slice is very pixelated and blurred, since the signal of the 1 mm slice is very weak (Fig. 19 a). If, however, averaging is performed based on several recorded MR images (b: 5, or c: 50), the quality increases very strongly (better signal-to-noise ratio, Figs. 19 b and c). In some cases, the identification of very fine structures, e.g. of a biological sample, requires a large number of averages. This is the only way to prevent the signal of the thin slice of a biological sample from getting lost in the noise and to ensure that the sample can be viewed in all its complexity and with all its intricacies.

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A: Generation of 2D MR images (7/16)

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A closer look at Fig. 19 c, for example, reveals that the central area of the MR image is very similar to the MR representation of pure oil (surroundings). This means that the selected 1 mm slice of the structure sample passes nearly completely through an opening of the plastic ladder. Only a very minor part of a ladder spoke is still included in the selected 1 mm slice, causing a slight darkening of the centre of the 2D MR image.

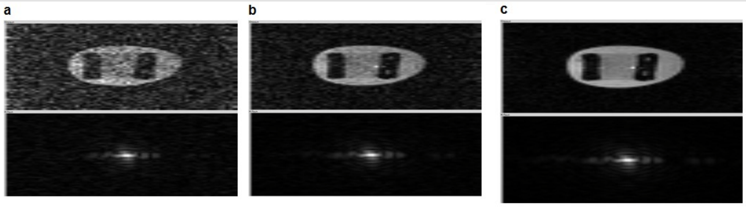

Fig. 19: Influence of the number of averages on the 2D MR image of a transverse 1 mm slice of the structure sample. In a, the transverse 1 mm slice of the structure sample was recorded with 1 average, in b with 5 averages, and in c with 50 averages. The settings of the essential parameters correspond to the first task. Obviously, the quality of the MR image of a very thin slice, i.e. the signal-to-noise ratio of this slice, increases when the number of averages increases.

A: Generation of 2D MR images (8/16)

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Generate a sagittal 2D MR image of the structure sample in oil in which the plastic ladder appears black. Apart from a specific position of the structure sample in the sample chamber, also a specific orientation of the "field of view" (FOV) must be selected.

Fig. 20 shows a sagittal 2D MR image of the structure sample in oil. The structure sample was aligned in the sample chamber so that the openings of the ladder face the back of the magnet. In this case, the two ladder beams made of plastic have different z-coordinates and all of the ladder spokes made of plastic are parallel to the z-axis. If the scanning window (see "Field of View") is now selected so that it includes the two ladder beams (alpha = 0°, beta = 90°), they will, of course, appear black in the MR image. The ladder spokes also appear black, since they are included in the scanning window if the angles that are mentioned above are used. The final result is a 2D MR image of a slice in which the plastic ladder appears black. The angle gamma = 90° flips the real 2D MR image once again by 90° and the image in Fig. 20 results. For the 2D MR image in Fig. 20, the parameters of task one were maintained to a large extent. Only the number of averages was increased to 10. At this point, we want to refer once again to the "field of view (FOV), which ultimately describes the size, position, and orientation of the scanning window in conjunction with the selected slice thickness.

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A: Generation of 2D MR images (9/16)

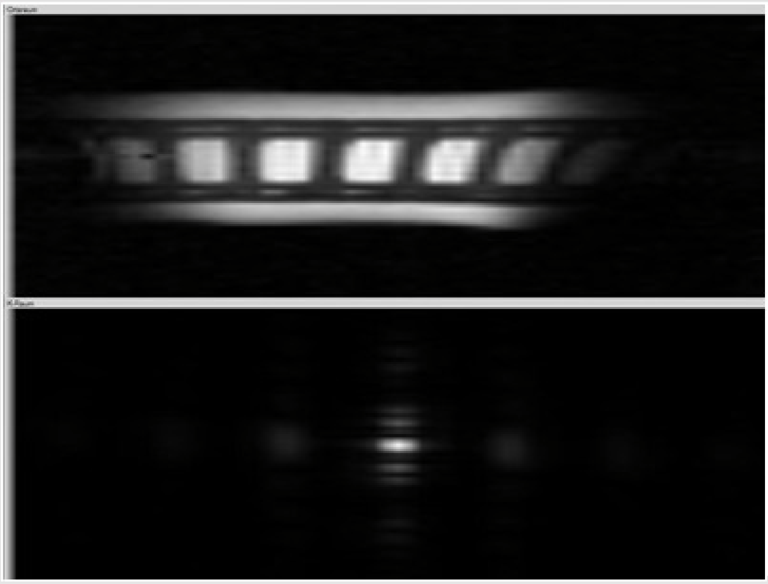

Fig. 20: Sagittal 2D MR image of the structure sample in oil. The openings of the ladder are aligned so that they are oriented towards the back of the magnet. The image has been recorded by way of a spatially encoded spin echo signal (Lo-calized Spin Echo 2D). The settings of the essential parameters correspond to the first task. Only the number of averages was increased to 10. In order to record the sagittal MR image, the parameters of the "Field of View" were set to the following values: FOV read: 20 mm, FOV phase: 20 mm, slice thickness: 10 mm, alpha: 0°, beta: 90°, gamma: 90°. The plastic ladder appears black, since the FOV includes the two ladder beams.

A: Generation of 2D MR images (10/16)

Then, generate the corresponding coronal 2D MR image while maintaining the alignment of the structure sample in the sample chamber (see point 4). Why does the ladder now appear bright? Answer this question by having a closer look at the structure sample.

Fig. 21 shows a coronal 2D MR image of the structure sample in oil. The structure sample was aligned in the sample chamber so that the openings of the ladder face the back of the magnet. In this case, the two ladder beams made of plastic have different z-coordinates is now selected so that it does not include the two ladder beams (alpha = 0°, beta = 0°), it is obvious that they cannot be reproduced in the MR image. Instead of the ladder beams, the oil will be reconstructed as a bright surface. The ladder spokes still appear black, since they are included in the scanning window if the angles that are mentioned above are used. The gaps between the ladder spokes are filled with oil so that they appear bright in the MR image. The overall result is a 2D MR image of a slice in which the existing oil forms a ladder-shaped structure in the 2D projection. This structure appears bright in the MR image. The angle gamma = 90° flips the real 2D MR image once again by 90° and the image in Fig. 21 results. For the 2D MR image in Fig. 21, the parameters of task one were maintained to a large extent. Only the number of averages was increased to 10.

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A: Generation of 2D MR images (11/16)

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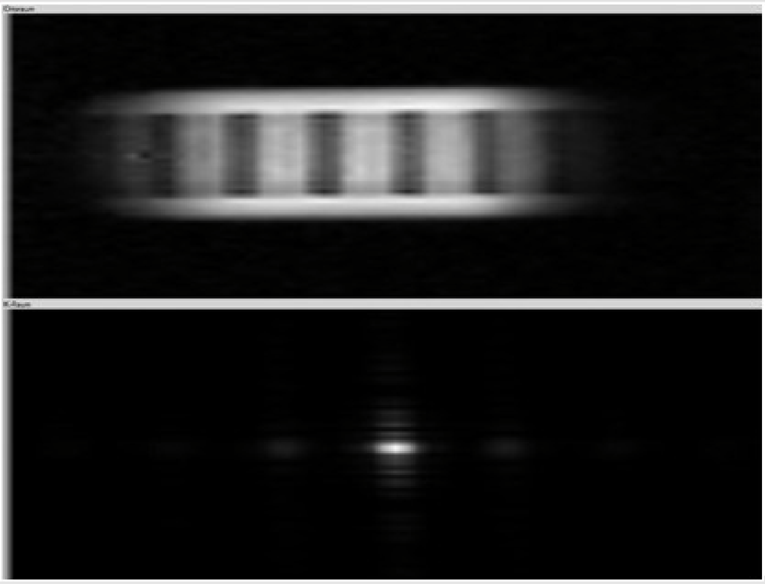

Fig. 21: Coronal 2D MR image of the structure sample in oil. The openings of the ladder are aligned so that they are oriented towards the back of the magnet. The image has been recorded by way of a spatially encoded spin echo signal (Localized Spin Echo 2D). The settings of the essential parameters correspond to the first task. Only the number of averages was increased to 10. In order to record the coronal MR im-age, the parameters of the "Field of View" were set to the following values: FOV read: 20 mm, FOV phase: 20 mm, slice thickness: 10 mm, alpha: 0°, beta: 0°, gamma: 90°. The plastic ladder appears bright, since the FOV does not include the two ladder beams.

A: Generation of 2D MR images (12/16)

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Following a 45° rotation of the structure sample in the sample chamber, try to generate an analogous 2D MR image like in task 1 by way of the setting of the orientation of the "field of view".

Fig. 22 a shows a transverse MR image of the structure sample in oil, which was recorded with the settings of task 1. Only the sample itself was rotated clockwise by 45° in the sample chamber prior to the recording of the image. Of course, the resulting MR image of the sample is also rotated by 45°. If the scanning window is also rotated clockwise by 45° now, i.e. if gamma is set to 135° instead of 90°, the resulting transverse MR image includes the structure sample with the same alignment as in task 1 (Fig. 22 b). What is decisive for the recording of an image is the alignment of the scanning window in relation to the alignment of the sample in the sample chamber. In order to produce the MR image in Fig. 22 b with a rotated sample in the sample chamber, other values for gamma are also possible. Of course, it is also possible to select $135^\circ + n \cdot 360^\circ$ with $n \in \mathbb{Z}$ for gamma. Due to the mirror symmetry of the sample, it is also possible to use the angles $135^\circ + n \cdot 180^\circ$ with $n \in \mathbb{Z}$.

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A: Generation of 2D MR images (13/16)

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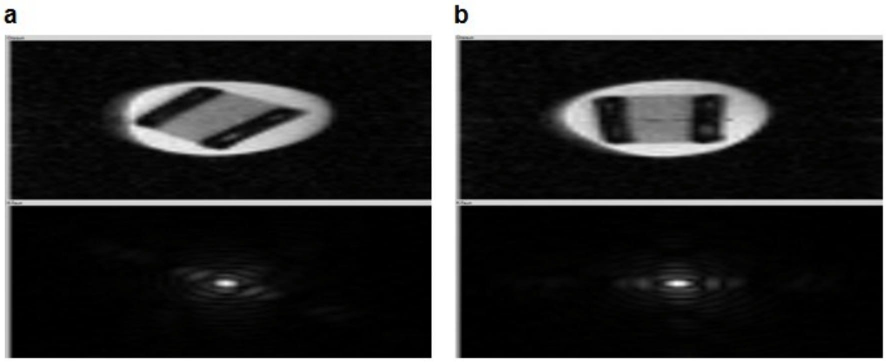

Fig. 22: Transverse MR images of the structure sample that were recorded with the settings of the first task. The sample itself was rotated clockwise by 45° in the sample chamber prior to the recording of the image. In a, the tilted transverse MR image of the structure sample results. In b, the scanning window was also rotated clockwise by 45°. In this case, the structure sample has the same alignment as the scanning window so that an MR image, like the one in task 1, results.

A: Generation of 2D MR images (14/16)

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Record 2D MR images of all kinds of samples with thin slice thicknesses. Beans, buds, and branches are particularly suitable. It is also possible to produce MR images of small insects with good resolution. Please note that, in the case of biological samples, the repetition time must be much longer than in the case of the structure sample. In addition, a very large number of averages is required for an adequate sectional MR image. This is why the total effective measuring time is clearly longer.

Fig. 23 shows the transverse MR image of a 1 mm slice of a thin pine twig. The following settings were used: data points = 128, phase steps = 128, repetition time = 750 ms, echo time = 12 ms, averages = 500, FOV read = 12 mm, FOV phase = 12 mm, alpha = 90°, beta = 0°, gamma = 90° (transverse slice). What catches the eye concerning the selection of the parameters is the relatively long repetition time and high number of averages. This is absolutely essential when recording the MR image of a thin slice of a biological sample. The necessity of a longer repetition time is based on the longer relaxation times in the case of biological samples. They are clearly longer than, for example, the relaxation time of oil. The high number of averages is necessary for an optimum signal-to-noise ratio (SNR) during the image recording of a very thin slice.

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A: Generation of 2D MR images (15/16)

The parameters that were selected above enable rather good levels of resolution. In Fig. 23, even the individual annual growth rings of the pine twig can be discerned. However, these types of images require a long measuring time (measurement over night).
One distinctive feature of the MR image in Fig. 23 should be explained in greater detail. In the middle of the image of the pine twig, there is a slightly thicker horizontal line that is clearly separated from the rest of the image. This is an image artefact that was caused by an imperfect 180° pulse. An imperfect 180° pulse evokes a small FID during its application. This "mini FID" is encoded and included in the image together with the other information and causes the line in the middle of the image. A so-called spoiler can minimise the influence of this type of image artefacts. In this case, two spoilers are applied symmetrically around the 180° pulse. The spoiler before the 180° pulse only dephases the magnetisation that results from the 90° pulse, while the spoiler after the 180° pulse dephases the total magnetisation after the 180° pulse. As a result, the dephasings concerning the evoked magnetisation of the 90° pulse cancel each other precisely out and the 180° pulse is brought to perfection. In addition, the 180° pulse can also be perfected by selecting a longer echo time. The spoiler settings can be found on the "Advanced" tab. Please note that although strong spoilers improve the 180° pulse, they also reduce the signal strength.

A: Generation of 2D MR images (16/16)

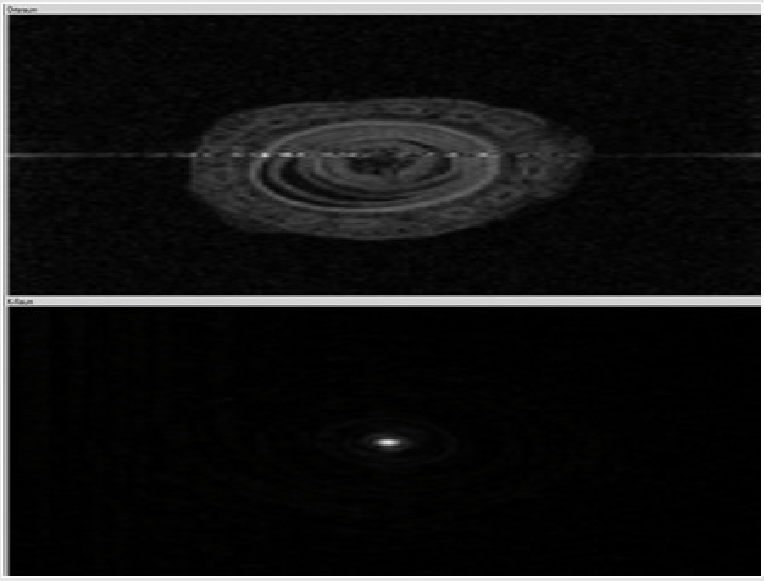

Fig. 23: Transverse MR image of a 1 mm slice of a thin pine twig. The following settings were used: data points = 128, phase steps = 128, repetition time = 750 ms, echo time = 12 ms, averages = 500, FOV read = 12 mm, FOV phase = 12 mm, alpha = 90°, beta = 0°, gamma = 90° (transverse slice). The individual annual growth rings of the pine twig are clearly visible. Please note, however, that, due to the long repetition time and high number of averages, the measuring time for producing the MR image had to be rather long (measurement over night).

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B: Generation of 3D MR images (1/10)

Record a 3D MR image of the structure sample with a comparatively small number of data points and phase steps (e.g. 16 in all of the three encoding directions).

Fig. 24 shows the 3D MR image of the structure sample with 16 data points in the first encoding direction, 16 phase steps in the second encoding direction, and 16 phase steps in the third encoding direction. The 3D image is a single image (number of averages = 1). It was recorded with a repetition time of 150 ms and an echo time of 10 ms. The two beams of the ladder (black), surrounded by oil (bright), are clearly visible. The slider on the right-hand side of the results panels can be used to obtain a more detailed interior view of the sample. The noise level (lower slider) has been adjusted so that the 3D sample is completely visible, but with as little noise as possible. In addition, the 3D image has been smoothed by way of an algorithm (see the settings in the "measure MRT" software).

B: Generation of 3D MR images (2/10)

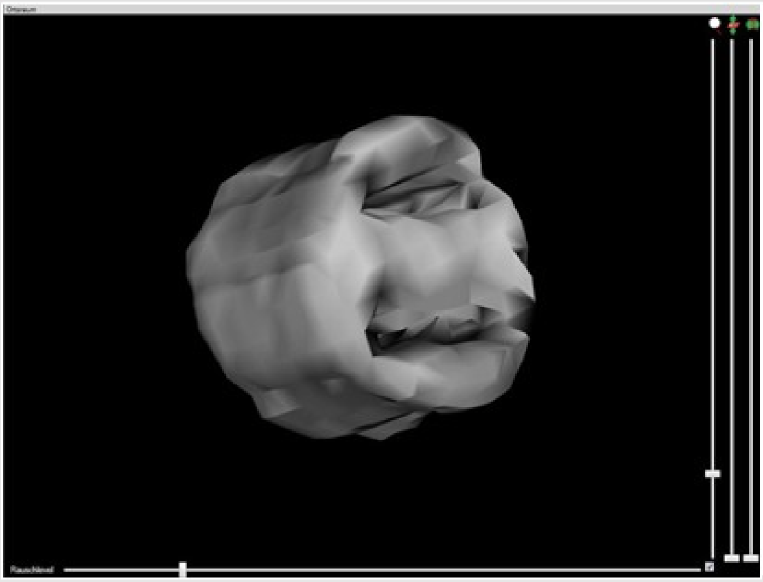

Fig. 24: 3D MR image of the structure sample, recorded with 16 data points in the first encoding direction, 16 phase steps in the second encoding direction, and 16 phase steps in the third encoding direction. The 3D image is a single image (number of averages = 1). It was recorded with a repetition time of 150 ms and an echo time of 10 ms. The ladder beams and partly also the ladder spokes are clearly visible (black) and they can be clearly distinguished from the surrounding oil (bright). The slider on the right-hand side of the results panels can be used to obtain a more detailed interior view of the sample. The noise level (lower slider) was optimised after the experiment.

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B: Generation of 3D MR images (3/10)

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Record a 3D MR image of the structure sample with a comparatively large number of data points and phase steps (e.g. 64 in all of the three encoding directions).

Fig. 25 shows the 3D MR image of the structure sample with 64 data points in the first encoding direction, 64 phase steps in the second encoding direction, and 64 phase steps in the third encoding direction. All of the other settings were adopted from task 1. In this case, too, the two beams of the ladder (black), surrounded by oil (bright) are clearly visible. Due to the very long measuring time for this type of 3D images, a higher number of phase steps is usually not useful. Recommendation: Set the data points and phase steps to 32 at maximum. This reduces the measuring time and, in most cases, leads to relatively good 3D images.

B: Generation of 3D MR images (4/10)

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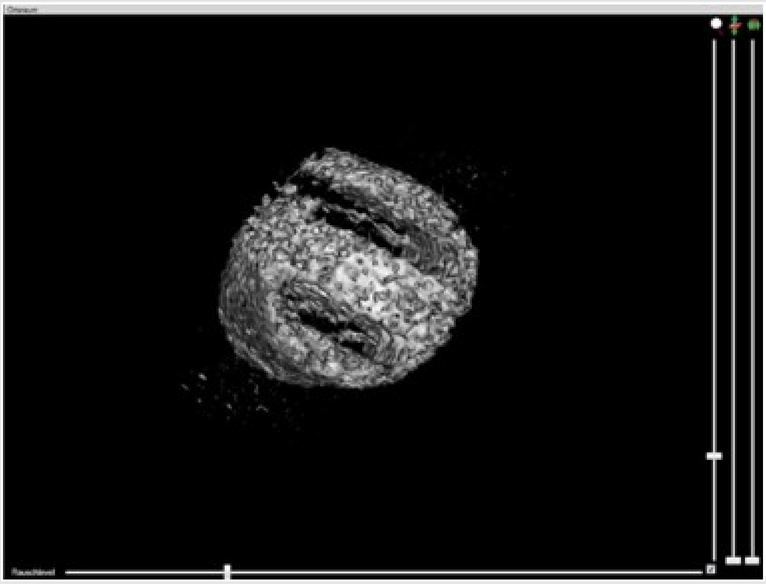

Fig. 25: 3D MR image of the structure sample, recorded with 64 data points in the first encoding direction, 64 phase steps in the second encoding direction, and 64 phase steps in the third encoding direction. All of the other settings were adopted from task 1. The ladder beams and partly also the ladder spokes are clearly visible (black) and they can be clearly distinguished from the sur-rounding oil (bright). This image required a long measuring time.

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B: Generation of 3D MR images (5/10)

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Examine the influence of the number of averages on the 3D MR image of the structure sample that was generated in 1.

Fig. 26 a shows a 3D MR image of the structure sample that was recorded with 3 averaging steps. Fig. 26 b shows this image, recorded with 10 averaging steps. The other settings were adopted from task 1 of this experiment part. It becomes clear that the 3D MR image of the structure sample hardly changes at all when the number of averaging steps increases. This means that the signal is already very strong after the first measurement. In this part of the experiment, a high number of averages has a considerably smaller effect than in the case of an MR representation of a thin slice.

B: Generation of 3D MR images (6/10)

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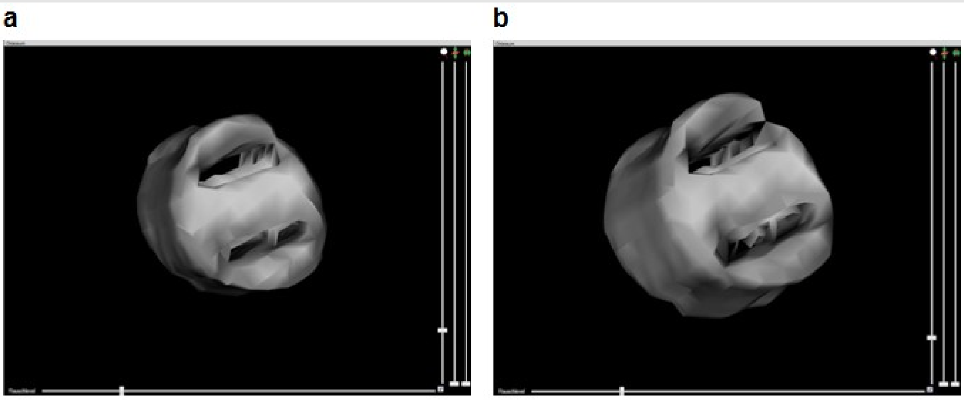

Fig. 26: 3D MR image of the structure sample, recorded with 3 averaging steps (a) and 10 averaging steps (b). The other settings were adopted from task 1 of this experiment part. It becomes clear that the 3D MR image of the structure sample hardly changes at all when the number of averaging steps increases. The signal-to-noise ratio is already comparatively high after only one averaging step.

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B: Generation of 3D MR images (7/10)

Record 3D MR images of all kinds of samples. Beans, buds, and branches are particularly suitable. Please note that, in the case of biological samples, the repetition time must be much longer than in the case of the structure sample. In addition, a larger number of averages is useful for obtaining an adequate MR image. This is why the total effective measuring time is clearly longer.

Fig. 27 shows the 3D MR image of the pine twig that has already been used in part A. The image was recorded with 32 data points in the first encoding direction, 32 phase steps in the second encoding direction, and 32 phase steps in the third encoding direction. The 3D image has been averaged over 10 individual measurements. It was recorded with a repetition time of 1 ms and an echo time of 12 ms. The noise level (lower slider) has been adjusted so that the 3D sample is completely visible, but with as little noise as possible. In addition, the 3D image has been smoothed by way of an algorithm (see the settings in the "measure MRT" software). In Fig. 27, the coarse structure of the pine twig with the bulges on the outside is clearly visible. The annual growth rings of the twig, however, can only be surmised. This is, of course, due to the lower resolution that had to be accepted in view of the shorter measuring time.

B: Generation of 3D MR images (8/10)

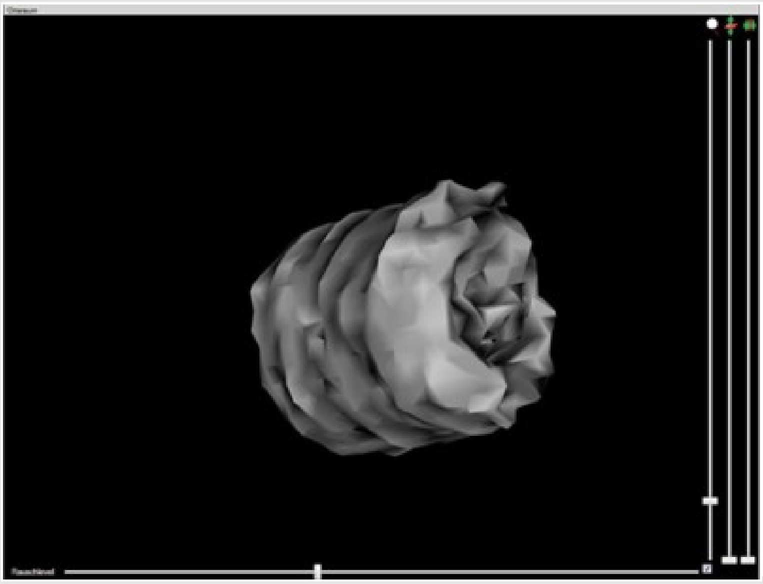

Fig. 27: 3D MR image of the pine twig that has already been used in part A. The im-age was recorded with 32 data points in the first encoding direction, 32 phase steps in the second encoding direction, and 32 phase steps in the third encoding direction. The 3D image has been aver-aged over 10 individual measurements. It was recorded with a repetition time of 1 ms and an echo time of 12 ms. With this rather low resolution, the course structure of the twig can be discerned, although not the individual growth rings. They are surmisable at best.

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B: Generation of 3D MR images (9/10)

Think about how the total measuring time for the generation of a 3D MR image is calculated. Take into consideration that the measuring time is mainly determined by the two phase encoding processes.

3D MR images require one frequency encoding and two phase encodings (see Fig. 16). Depending on the resolution, several phase encoding steps are required in the two phase encoding directions. Let us look at the example of a cube of 4 x 4 x 4 voxels. One slice of the cube requires 4 different phase encoding steps of the regular phase encoding gradient. For the recording of all 4 slices, 4 additional phase encoding steps are required for the additional phase encoding gradient. If the resolution is doubled in one of the two phase encoding directions, the measuring time will also be doubled (see "Spatial encoding in nuclear magnetic resonance"). However, if the resolution is doubled in the frequency encoding direction, the measuring time will not be doubled (see "Spatial encoding in nuclear magnetic resonance").

B: Generation of 3D MR images (10/10)

The assignment of the corresponding grey values to a cube of 8 x 8 x 8 voxels requires 4 times the measuring time as a cube of 4 x 4 x 4 voxels. In summary, it can be said that the doubling of the resolution in all 3 spatial directions requires a 4 times longer measuring time. If the selection of a longer repetition time and a higher number of averages are also taken into consideration, it becomes clear rather quickly that the recording of 3D MR images can easily take several hours. This can be verified simply by varying the parameters directly with the "measure MRT" software, since the calculated measuring time is always indicated in the parameters area.

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