Spatial encoding in Nuclear Magnetic Resonance

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

This content can also be found online at:

http://localhost:1337/c/5fdcace85098f00003f1f037

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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 a tool for imaging.

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Other information (1/2)

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Prior knowledge

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

Main principle

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 1D spatial encoding by way of an additional magnetic gradient field in the encoding direction and the visualisation of spatiotemporal T_1 and T_2 profiles. T_1 and T_2 are specific for the sample material that is to be analysed and, thereby, provide important information concerning its characteristic composition.

Other information (2/2)

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Learning objective

The aim of these experiments is to study how spatial encoding works in MR technology.

1. Study the effects of a magnetic gradient field that is superimposed on the external static magnetic field $\vec{B}_0$ on the spatially encoded representation of a combined oil and water sample (double sample).
2. Record an ideal profile of the double sample, i.e. display the signals of oil and water in a spatially resolved manner. Both signals must be separately visible. Observe this profile at different repetition times.
3. Study the behaviour of an ideal profile of the double sample when this double sample is rotated in the sample chamber (90°, 180°).
4. Repeat for the spatiotemporal T_1 and T_2 profil.

Tasks

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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 (1/27)

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The starting point for the theory concerning this experiment ensemble is the slightly preferred parallel alignment of nuclear spins in a static magnetic field $\vec{B}_0$ and the associated generation of 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:

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

A 90° HF pulse in the resonance condition (1) that is applied perpendicularly to the external static magnetic field $\vec{B}_0$ deflects the total magnetisation by 90°, i.e. the initial longitudinal magnetisation $\overrightarrow{M_{L0}}$ is completely transformed into a transverse magnetisation $\overrightarrow{M_Q(0)}$ that then precesses around the static magnetic field vector with the Larmor frequency. Every deflection means an abandonment of the state of equilibrium. The relaxation describes the natural, dynamic restoration of the original state of equilibrium.

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Theory (2/27)

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The exponential restoration of the longitudinal magnetisation $\overrightarrow{M_L(t)}$ is described by the relaxation time T_1, while 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 (3/27)

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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). Based on various interactions, T_2 is normally smaller than T_1. However, up to this point, it is still unclear as to how an MR image is actually generated, i.e. how slices are selected and how signals are provided with a spatial assignment. The following will show how spatial signals can actually be distinguished.

The trick is relatively simple, yet incredibly effective. We apply position-dependent magnetic gradient fields in addition to the external static magnetic field $\overrightarrow{B_0}$. These gradients are usually generated by opposite pairs of coils in the x-, y-, and z-direction. These coils are operated at the same current intensity, but with opposite polarity. As a result, the static magnetic field is increased at one coil and decreased at the opposite coil (see Fig. 1). The effect is that the individual nuclear spins precess with different speeds at different locations, i.e. they show resonance at different frequencies.

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Theory (4/27)

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If, for example, a magnetic and linear gradient field is applied simultaneously with the HF excitation pulse, a specific slice can be selected (slice selection gradient). Let us assume that the gradient is applied symmetrically around z = 0 in the z-direction. In this case, the following applies to the changed static magnetic field in terms of its absolute value:

$$B_z(z) = B_0(0) + G_z \cdot z \qquad (4)$$

with G_z as the strength of the change of the magnetic field per length in the z-direction (gradient) (see Fig. 2).

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. The thickness of the slice can be selected in two different ways. Firstly, the HF pulse can be provided with a larger bandwidth. In this case, the neighbouring frequencies would also excite spins in the range of z_0 and bring them into resonance. Secondly, it is also possible to leave the bandwidth as it is and to change the strength of the magnetic gradient field. A weaker gradient brings a thicker slice into resonance, while a stronger gradient brings a correspondingly thinner slice into resonance (see Fig. 2).

Theory (5/27)

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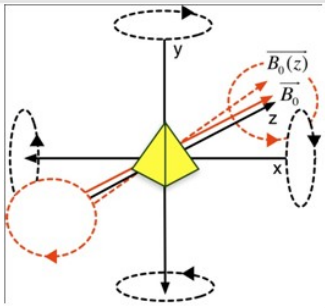

Fig. 1: Generation of magnetic gradient fields by way of opposite pairs of coils with the same current intensity but different polarity. The nuclear spins are excited by the HF pulse with the Larmor frequency ω_0 only at a certain resonance point z_0. This is the slice position that is to be examined.

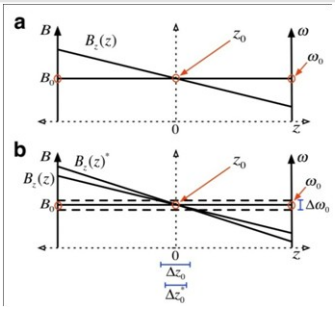

Fig. 2: Gradient field for the slice selection in the z-direction. Figure (a) shows that, due to the position-dependent gradient or due to the new, position-dependent magnetic field B_z (z), the nuclear spins come into resonance with the HF pulse with the frequency ω_0 only at the point z_0. Figure (b) 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 (6/27)

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It is clear that, due to three pairs of gradient coils, it is possible to select any slice planes in space. The three main slice planes have different names in MR diagnostics. Transverse slices are sections from head to foot, sagittal slices are sections from the right arm to the left arm, and coronal slices are sections from the nose to the back of the head. Up to this moment, the gradient field has only been used for the slice selection. Now, the question is how an MR image of a slice can be generated, i.e. how the signals can be assigned to their exact location of origin. Every MR image consists of individual image elements, the so-called pixels (2D) or voxels (3D) (from this point on, we will use the term voxel in order to account for the slice thickness). These individual elements have characteristic grey values that are proportional to the signal strength. The resolution of an image is simply characterised by the number of voxels. This means that the higher the number of voxels is, the higher also the number of different grey values is and the more pieces of signal information must be assigned to their location of origin. As a result, an MR image that consists of 64 (height) x 64 (width) voxels requires 4096 individual signals to be distinguished (see Fig. 3). Once again, magnetic gradient fields are applied for the discrimination of these individual signals and the final assignment to their respective locations of origin. If applied in a clever manner, this method enables the enforcement of spatial encoding in nuclear magnetic resonance (see Fig. 3).

Theory (7/27)

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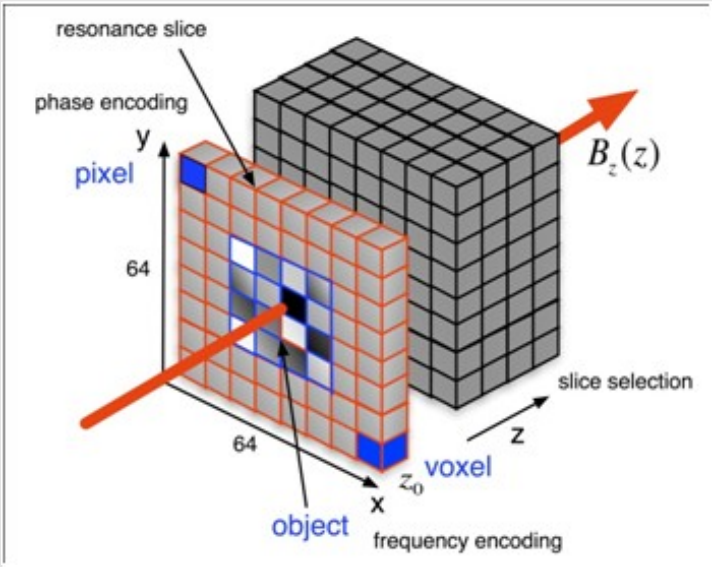

Fig. 3: Voxel representation of the slice selection. Based on an additional gradient field in the z-direction, only nu-clear spins in the resonance slice will be excited. The slice thus selected is spatially encoded in the x- and y-direction by way of additional gradient fields (e.g. frequency encoding in the x-direction and phase encoding in the y-direction).

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Theory (8/27)

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Now, the first step is to demonstrate 1D spatial encoding, i.e. the spatial encoding of a voxel strip. For this purpose, we look at our spin echo signal that has already been presented in detail in the experiment ensembles "Fundamental principles of nuclear magnetic resonance" and "Relaxation times in nuclear magnetic resonance". The main advantages of the echo signal compared to the direct FID signal are based on the clearness of the measurement of the maximum amplitude and the gain of time involved in a spin echo sequence that enables the application of information to the system prior to the signal analysis. In addition, the relaxation time T_2 of a medium can be read directly based on the spin echo amplitude. 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 voxel of this strip, like a sound that includes numerous different frequencies.

Theory (9/27)

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A magnetic gradient thus applied is called a frequency encoding gradient and the corresponding method is referred to as frequency encoding (see Figs. 4 and 6). The trick is that the signal strength of a certain voxel can be assigned precisely to this voxel based on the frequency. The result is a projection of the medium that is to be studied on the gradient axis (see Fig. 4). The signal strength determines the grey value of the associated voxel. The mathematical method that ensures this assignment is the so-called Fourier transformation. Its effect is similar to that of a prism that breaks up light into its spectral colours. The process that is described above can be used for the relatively quick performance of the 1D spatial encoding of a medium, i.e. frequency encoding can be used to create and analyse the 1D profiles of a sample. 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 (10/27)

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If we briefly apply a linear magnetic field gradient 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 precess briefly 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 Figs. 5 and 6). However, this is not sufficient yet, since a dispersion into position-specific frequency components via a Fourier transformation cannot be performed based on one single phase data point. If the measurement is repeated with a different duration or amplitude (spin warp imaging; preferred method) of the linear magnetic field gradient, the signal course of the various voxels can be reconstructed in an unambiguous manner. The trick is that the Fourier transformation enables the individual 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. 5).

Theory (11/27)

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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. This means that for the 2D MR imaging of a matrix of 64 x 64 voxels, we need to repeat the pulse sequence that is described above 64 times. It now becomes clear that the duration of an MR scan is mainly determined by the phase encoding process (see Figs. 5 and 6).

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Theory (12/27)

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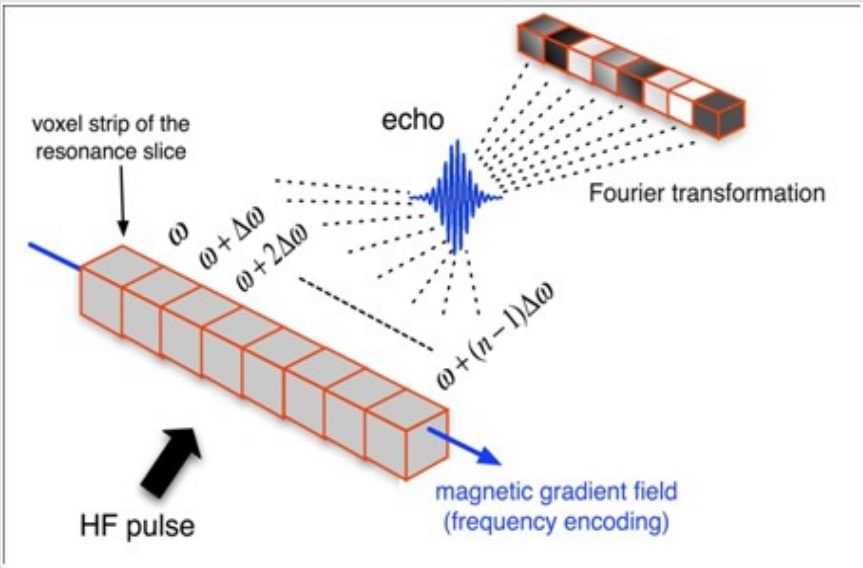

Fig. 4: Spatial encoding of a 1D voxel strip (n voxels). During the measurement of the spin echo, a linear magnetic field gradient is applied in the encoding direction. This gradient causes the nuclear spins of the voxel strip to precess with a frequency that increases linearly (frequency encoding). 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 (13/27)

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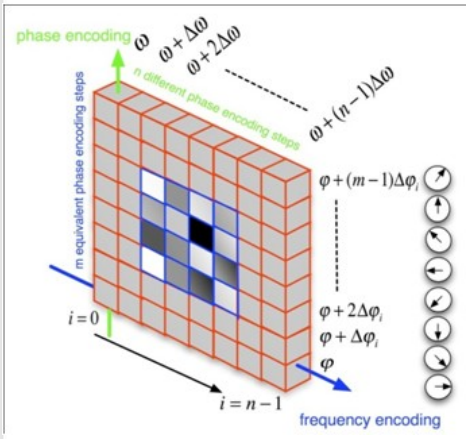

Fig. 5: Spatial encoding of a 2D voxel slice (n x m voxels). During the measurement of the spin echo, a linear magnetic field gradient is applied in one of the two encoding directions (e.g. in the x-direction). This gradient causes the nuclear spins along this direction to precess with a fre-quency that 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 (14/27)

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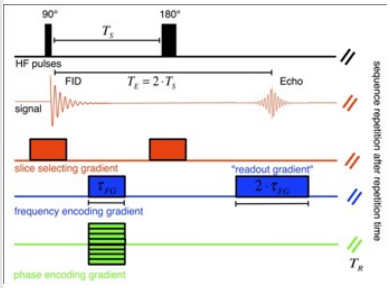

Fig. 6: Temporal application of three magnetic gradient fields for the 2D imaging of a slice. During the two HF pulses (90°, 180°), the "slice selection gradient" is applied. This gradient brings only the nuclear spins of a certain slice with a corresponding slice thickness into resonance (red). During 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 (15/27)

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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 well-defined direction. As a result, every point corresponds to a stripe pattern. 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 strip 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 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. 7). Every point in the k-space includes information concerning the entire image in the position space, i.e. the mapping from k-space to position space is not injective. Data points in the centre of the k-space determine the signal-to-noise ratio, structure, and contrast of the reconstructed image. Outer data points provide information concerning the borders, edges, contours, and fine transitions. Thereby, it also becomes clear that the dimensions of a voxel are inversely proportional to the maximum k_x and k_y value (see Fig. 7).

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Theory (16/27)

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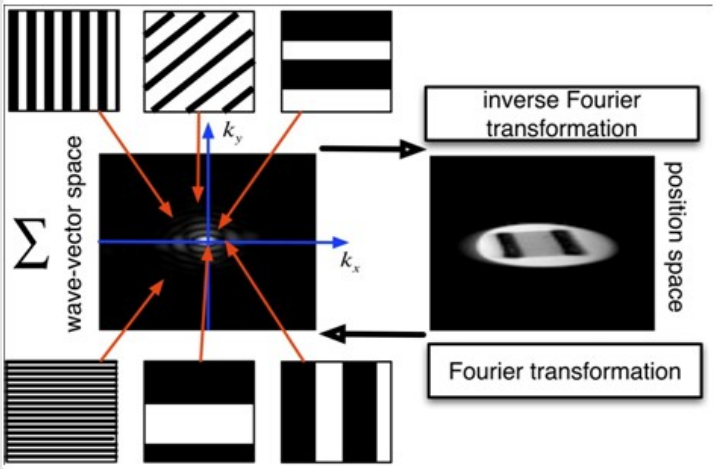

Fig. 7: 2D imaging in the k-space and position space. In the k-space, every point corre-sponds to a stripe pattern with a well-defined frequency. The raw data values in the k-space determine the weighting of these stripe patterns and, thereby, the wave-vector image. Coarser strip patterns are very close 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 (17/27)

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2D imaging is discussed in greater detail in the experiment ensemble "Magnetic resonance imaging I". Let us now return to the 1D imaging, i.e. to the imaging with a frequency encoding gradient. So far, our sequence for the visualisation of a profile in the encoding direction consists of a 90° HF excitation pulse that generates the FID signal and a 180° pulse after the time T_S that causes the nuclear spin ensemble to rephase in the time $T_E = 2 \cdot T_S$. The resulting spin echo signal is superimposed by the magnetic gradient field for spatial encoding ("readout gradient"). Based on the resulting signal mixture, the frequency profile is reconstructed by way of a Fourier transformation. The contribution of a frequency to the signal mixture is assigned to this specific frequency (see Fig. 8).

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Theory (18/27)

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The problem is that, during the frequency encoding of the spin echo signal, the precession of the individual nuclear spins in the direction of the readout gradient fans out systematically, which is undesired. As a result, the nuclear spin ensemble would have already dephased at the point of time of the echo T_E, which means that no echo signal could be measured. The trick is to enforce the artificial dephasing of the nuclear spin ensemble prior to the frequency encoding. For this purpose, a so-called "dephasing gradient" is used. The aim of this gradient is the dephasing of the nuclear spin ensemble in the opposite direction than the one that is evoked by the readout gradient. In order to ensure that the nuclear spin ensemble has rephased at the point of time of the echo T_E and that the spin echo signal reaches its maximum amplitude at this point of time, the duration of the dephasing gradient must be half the duration of the readout gradient. If the dephasing gradient is applied after the 180° pulse, it must have the opposite polarity than the readout gradient. In most cases, however, this gradient is applied prior to the 180° pulse. Since a 180° pulse reverses the polarity, the dephasing gradient must have the same polarity as the readout gradient in this case (see Fig. 8).

Theory (19/27)

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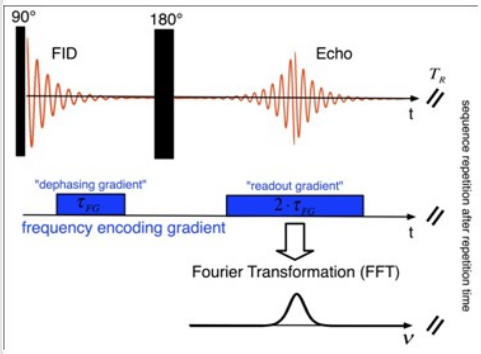

Fig. 8: Sequence representation for the generation of a 1D profile of a sample. During the spin echo, a readout gradient of the duration $2\tau_{FG}$ is applied. It ensures the frequency encoding of the signals of the voxel strip. An FFT (fast Fourier transformation) can be used to decompose the resulting signal mixture into its individual frequencies and to represent it in a spatially resolved manner. During frequency encoding, there is usually an artificial fanning-out of the nuclear spins. This is counteracted by way of a "dephasing gradient" of half the duration τ_{FG} of the "readout gradient". The repetition time T_R between the individual measurements is crucial for the contrast between different substances of a sample. Substances with a short relaxation time show nearly the full signal strength in the subsequent measurement already at short repetition times T_R. In the case of substances with very long relaxation times, the signal strength in the subsequent measurement decreases considerably in the case of a short repetition time T_R.

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Theory (20/27)

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The question is how a spatially encoded profile of a sample can be provided with signal weighting based on the substance-specific relaxation times T_1 or T_2. This weighting is an essential part of all MR images, since it ensures the levels of contrast that are necessary to distinguish cerebrospinal fluid from fat, water from oil, tumours from healthy tissue, etc. The profile representation sequence that is presented here is already sufficient for visualising different substances. This can be achieved by varying the repetition time between several individual measurements with the same sequence. If this time is short in the case of substances with a long relaxation time, the deflected magnetisation vector does not return completely to the state of equilibrium, i.e. the parallel position with regard to the external static magnetic field $\vec{B}_0$. As a result, the recorded spin echo signal will be weaker in the subsequent measurement. However, this method is not suitable for an adequate relaxation time weighting.
An adequate T_1 relaxation time weighting during a measurement requires a sequence of three pulses. The first 90° pulse deflects the magnetisation vector into the plane perpendicular to $\vec{B}_0$. The longitudinal magnetisation $\overrightarrow{M_{L0}(t)}$ is transformed into a transverse magnetisation $\overrightarrow{M_Q(0)}$.

Theory (21/27)

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After a variable time $\tau_{\Delta 90^\circ}$, a second 90° pulse is applied. The resulting FID signal is proportional to the longitudinal magnetisation $\overrightarrow{M_L(\tau_{\Delta 90})}$ that is restored after the time $\tau_{\Delta 90^\circ}$ (see "Relaxation times in nuclear magnetic resonance"). Then, the dephasing gradient is applied again before a 180° pulse after the time T_S (measured after the second 90° pulse) evokes a spin echo signal after the time $T_E = 2 \cdot T_S$ (measured after the second 90° pulse). The spin echo signal is superimposed by the readout gradient. A Fourier transformation of this signal leads to a profile of the sample. This profile includes a signal weighting based on the substance-specific relaxation time T_1 (see Fig. 9). If we start with a small $\tau_{\Delta 90^\circ}$ and if we increase this value in several consecutive measurements, the result is a spatiotemporal T_1 profile of the sample. The repetition time \VT_R\) between the measurements during the spatiotemporal T_1 profile measurement is crucial for the strength of the signal of a specific sample that is to be analysed. In the case of samples with short relaxation times, a relatively short repetition time is usually sufficient for obtaining a comparably strong signal in the subsequent measurement (e.g. oil). In this case, the magnetisation vector has reached its ground state before the next measurement commences.

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Theory (22/27)

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In the case of samples with longer relaxation times, the repetition time must be extended accordingly in order to produce adequate signal strength in the subsequent measurement (e.g. water). Insufficient repetition times would lead to disequilibrium at the beginning of the subsequent measurement. This means that, after the first 90° pulse, the magnetisation vector would not correspond to a complete transverse magnetisation (see Fig. 9). Thereby, it becomes clear that the repetition time can be used to adjust the contrast of the profile of a double sample of oil and water, for example. A relatively strong contrast can be reached in the case of repetition times between the T_1 relaxation times of oil and water.

Theory (23/27)

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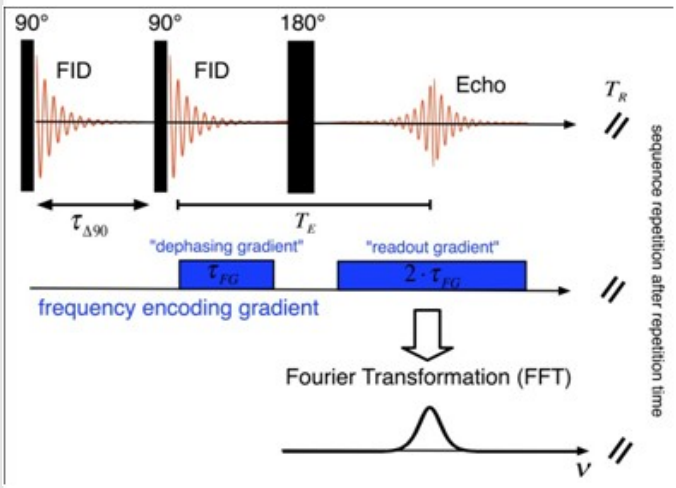

Fig. 9: Sequence representation for the generation of a 1D profile of a sample with T_1 relaxation time weighting. This measurement requires a sequence of 3 HF pulses. The first 90° pulse deflects the magnetisation vector into the plane that is perpendicular to the static magnetic field (B_0). A second 90° pulse after a variable time $\tau_{\Delta 90^\circ}$ measures the absolute value of the restored portion of the longitudinal magnetisation $M_L(\tau_{\Delta 90^\circ})$ after precisely this time $\tau_{\Delta 90^\circ}$. The resulting FID signal is then further processed as shown in Fig. 8. The FFT of the spin echo signal mixture leads to the spatially resolved representation of a T_1-weighted 1D profile of a sample. If the individual measurement is repeated with a varying $\tau_{\Delta 90^\circ}$, a spatiotemporal and T_1-weighted 1D profile of a sample can be generated.

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Theory (24/27)

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The exact T_2 relaxation time weighting during a measurement requires the same sequence that has already been used for the profile representation. The first 90° pulse deflects the magnetisation vector into the plane perpendicular to $\overrightarrow{B_0}$. The longitudinal magnetisation $\overrightarrow{M_{L0}(t)}$ is transformed into a transverse magnetisation $\overrightarrow{M_Q(0)}$ that dephases after the time T_2. The result is the classic FID signal. Then, the dephasing gradient is applied again in order to let the signal fan out in the opposite direction of the one that is evoked by the later readout gradient. After a variable time T_S, a 180° pulse is applied. This pulse generates the spin echo signal after the time \T_E = \tau_{\Delta 180} = 2 \cdot T_S\). For spatial encoding, the spin echo signal is then superimposed by the readout gradient (see Fig. 10). A Fourier transformation of this signal leads to a profile of the sample. This profile includes a signal weighting based on the substance-specific relaxation time T_2. If we start with a small \T_E = \tau_{\Delta 180}\) and if we increase this value in several consecutive measurements, the result is a spatiotemporal T_2 profile of the sample (see Fig. 10). For a T_2 relaxation time weighting of the profile of a double sample of oil and water, for example, longer repetition times than for the T_1 relaxation time weighting are usually used.

Theory (25/27)

PHYWE

However, the sequence for the measurement of a T_2 profile has a disadvantage. After the first excitation with a 90° pulse, a certain waiting time T_S passes before the second 180° pulse evokes the rephasing of the spin ensemble. For the measurement of a complete and exact T_2 profile, it would be important to also permit rather long application times T_S or at least application times that are within the range of the T_2 relaxation time of the medium that is analysed. However, in the case of long application times T_S, the signal dephases to such an extent that the nuclear spin ensemble cannot rephase at all or only in an incomplete manner. This is due to the fact that the inhomogeneities of the magnet and the diffusions in the case of long application times (waiting times) T_S increasingly gain influence and evoke a dephasing of the spin ensemble that cannot be undone by way of a spin echo signal. As a result, during the above-mentioned measurement of the T_2profile, the signal decreases more quickly than with the T_2 relaxation time of the medium that is analysed.

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Theory (26/27)

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Please note that this phenomenon can also be observed during a relaxation time measurement with a multi-echo sequence (see "Relaxation times in nuclear magnetic resonance"). In order to be able to measure the exact T_2 relaxation time of a substance, the multi-echo sequence that is described in the experiment ensemble "Relaxation times in nuclear magnetic resonance" uses echo times that are as short as possible, since in this case every echo quickly rephases the nuclear spin ensemble so that unwanted dephasing processes are avoided. If, on the other hand, the echo times are extended, the unwanted dephasing processes occur. In this case, the measurement of the spin echo amplitudes leads to a T_2 relaxation time that is shorter than the true T_2 relaxation time of the analysed sample.

Theory (27/27)

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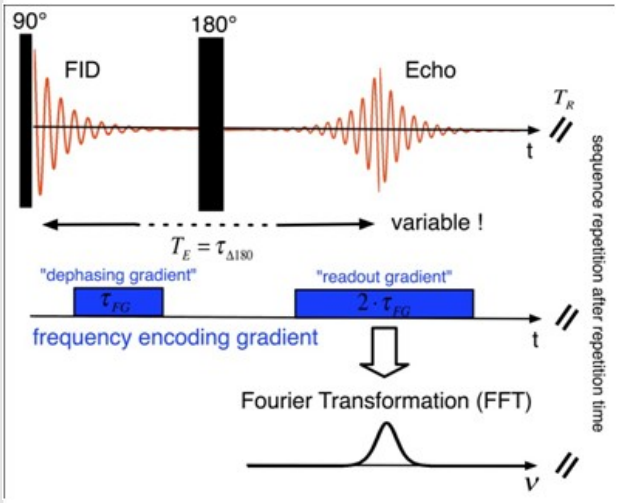

Fig. 10: Sequence representation for the generation of a 1D profile of a sample with T_2 relaxation time weighting. This measurement requires a sequence of 2 HF pulses. The first 90° pulse deflects the magnetisation vector into the plane that is perpendicular to the static magnetic field (B_0). A second 180° pulse after a variable time T_S generates a spin echo signal after a variable time $\tau_{\Delta 180}$. The amplitude of the spin echo corresponds to the absolute reduced portion of the transverse magnetisation $(M_Q(\tau_{\Delta 180})$ after the time $\tau_{\Delta 180}$. The spin echo signal is then further processed as shown in Fig. 8. The FFT of the spin echo signal mixture leads to the spatially resolved representation of a T_2-weighted 1D profile of a sample. If the individual measurement is repeated with a varying $\tau_{\Delta 180}$, a spatiotemporal T_2-weighted 1D profile of a sample can be generated.

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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)

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)

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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. 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

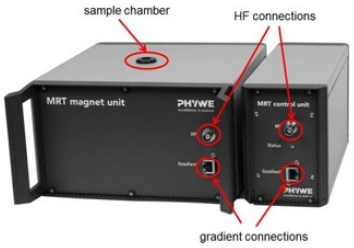

Fig. 13: Connectors at the back of the control unit

Procedure (1/8)

PHYWE

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".

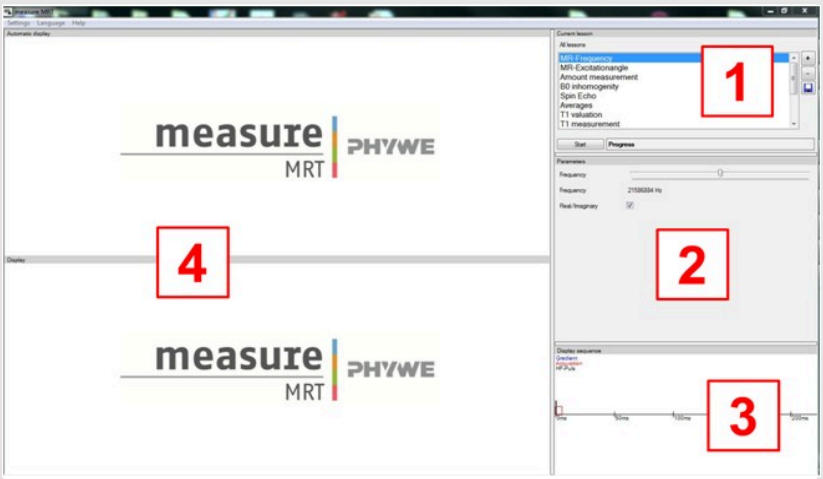

Fig. 14: Areas of the "measure MRT" program

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Procedure (2/8)

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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 (3/8)

PHYWE

A: 1D spatial encoding by way of a magnetic gradient field (frequency encoding)

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 Profile. The parameters area now shows the setting options Repetition time, Gradient X, Gradient Y, and Gradient Z (see Fig. 15). Set the repetition time to 5 s. Vary the gradient magnetic field that is superimposed on the static magnetic field by way of the sliders Gradient X,Y, and Z.

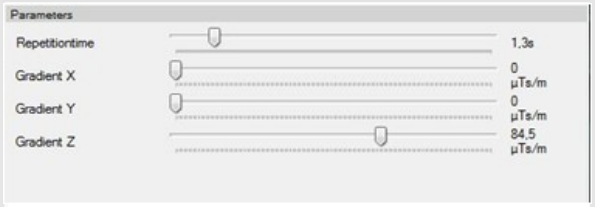

Fig. 15: Profile - parameters

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Procedure (4/8)

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2. Adjust the gradient by way of the sliders Gradient X, Gradient Y, and Gradient Z until you obtain a clear profile of the double sample, i.e. until the oil and water samples are visible in a separate manner. If the samples are aligned in parallel to the rear edge of the magnet housing in accordance with 1, a specific gradient slider must be readjusted to ensure that the oil and water samples can actually be distinguished from one another. This enables conclusions to be made concerning the arrangement of the gradient coils in the MR unit. The strength of the gradient field determines the profile.
   Then, ensure to also vary the repetition time by way of the slider Repetition time. A very long repetition time makes oil and water seem nearly identical.

3. Rotate the double sample in the MR sample chamber by 90° and 180° while the repetition time is comparably short (e.g. 1 s, discernibility of oil and water). Observe the profile after these rotations.

Procedure (5/8)

PHYWE

B: Visualisation of a spatiotemporal T_1 profile in the encoding direction

1. Place the oil and water double sample into the sample chamber in accordance with part A. In the experiments area (lessons), select the lesson T1 profile. The parameters area now shows the setting options Repetition time, Gradient X, Gradient Y, Gradient Z, Number of points, and Time step (see Fig. 16). Adjust the Repetition time, Gradient X, Gradient Y, and Gradient Z in the same manner as in part A so that an ideal profile of the double sample can be recorded. Vary the time step and the number of profiles that are to be recorded by way of the sliders Time step and Number of points.

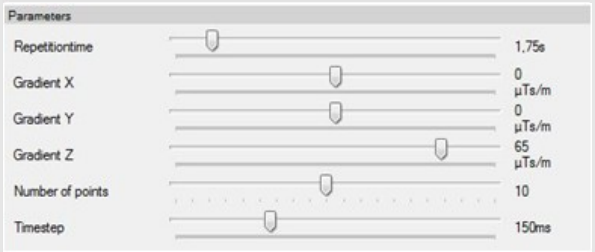

Fig. 16: T1 profile - parameters

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Procedure (6/8)

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2. Proceed in the same manner as for the first part and set the repetition time to approximately 1 second. Then, adjust the time step and number of points so that a good spatiotemporal visualisation of the T_1 profile can be achieved. The pulse duration of the last measurement should be longer than the estimated T_1 of water (see "Relaxation times in nuclear magnetic resonance" P5942200). A large number of measurements (number of points) is always useful for a clear visualisation.

Procedure (7/8)

PHYWE

C: Visualisation of a spatiotemporal T_2 profile in the encoding direction

1. Place the oil and water double sample into the sample chamber in accordance with part A. In the experiments area (lessons), select the lesson T2 profile. The parameters area now shows the setting options Repetition time, Gradient X, Gradient Y, Gradient Z, Number of points, and Time step (see Fig. 17). Adjust the Repetition time, Gradient X, Gradient Y, and Gradient Z in the same manner as in part A so that an ideal profile of the double sample can be recorded. Vary the time step and the number of profiles that are to be recorded by way of the sliders Time step and Number of points.

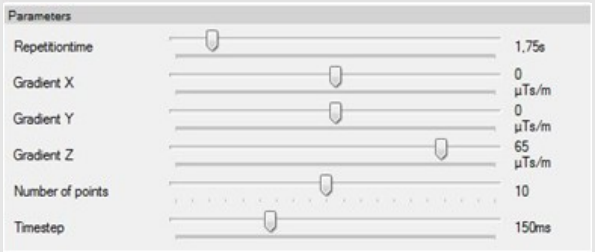

Fig. 17: T2 profile - parameters

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Procedure (8/8)

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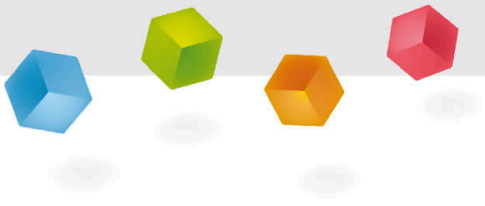
2. Proceed in the same manner as for the first part. Set the repetition time to approximately 2 seconds. Then, adjust the time step and number of points so that a spatiotemporal visualisation of the T_2 profile results. Due to the special sequence, the spin ensemble dephases very quickly in the case of long time steps. This is why you should select a very short time step and record the strongly shortened spatiotemporal T_2 profile of the double sample (it is not possible to record an exact spatiotemporal T_2 profile with the measurement sequence that is used here).

PHYWE



Evaluation

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A: 1D spatial encoding by frequency encoding (1/8)

PHYWE

Study the effects of a magnetic gradient field that is superimposed on the external static magnetic field $\vec{B_0}$ on the spatially encoded representation of a combined oil and water sample (double sample).

Figs. 18 a-c show the spatially encoded representation of a double sample for three exemplary gradient fields. The double sample is aligned in parallel to the rear edge of the magnet housing. Fig. 18 a shows the spatially encoded representation in the case of a gradient of the strength 75.4 $\mu Ts/m$ that is applied in the x-direction. Fig. 18 b shows the spatially encoded representation in the case of a gradient of the strength 75.4 $\mu Ts/m$ that is applied in the y-direction. Fig. 18 c shows the spatially encoded representation in the case of a gradient of the strength 75.4 $\mu Ts/m$ that is applied in the z-direction. Please note that all of the spatially encoded representations result from a summation of the signal over the entire sample space, since the sequence that is described in Fig. 8 does not involve any slice selection.

A: 1D spatial encoding by frequency encoding (2/8)

PHYWE

In Figs. 18 a and 18 b, the individual signals of the oil and water sample cannot be distinguished. This is immediately clear based on the special position of the double sample and when considering the well-defined gradient direction (compare the theory). The gradient in the x-direction goes from the front of the magnet housing to its back. This means that it is perpendicular to the specially aligned double sample, i.e. the gradient is perpendicular to the parting line between the oil and water samples. The gradient in the y-direction goes from the bottom of the magnet housing to its top, which means that it is also perpendicular to the specially aligned double sample. As a result, both gradients supply a spatially encoded and summed-up "joint signal" of both samples, since every slice of the sample space that is orthogonal to the x- or y-direction includes no sample at all or both samples (oil, water). The situation is different as far as the gradient in the z-direction is concerned. This gradient is parallel to the specially aligned double sample and, thereby, provides a spatially encoded and summed-up signal in which every sample is visible in a separate manner (ideal profile of the double sample). This is due to the fact that every slice of the sample space that is orthogonal to the z-direction never includes both samples (oil, water) at the same time.

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A: 1D spatial encoding by frequency encoding (3/8)

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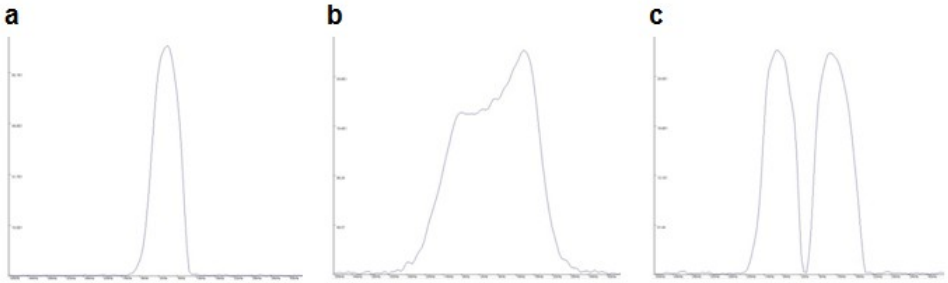

Fig. 18: Spatially encoded representation of a double sample of oil and water for three exemplary gradient fields. The double sample is aligned in parallel to the rear edge of the magnet housing. A gradient of the strength 75.4 μTs/m is applied (a) in the x-direction and (b) in the y-direction. In this case, the individual signals of the oil and water sample cannot be distinguished, since both gradients are orthogonal with regard to the parting line between the oil and water samples. In (c), however, the gradient with the strength of 75.4 μTs/m in the z-direction is parallel to the parting line. In this case, the profile of the double sample can be clearly decomposed into two separate individual signals. One can be assigned to the oil sample and the other to the water sample.

A: 1D spatial encoding by frequency encoding (4/8)

PHYWE

Figs. 19 a-c show the ideal profile of the double sample for an increasing gradient strength in the z-direction (a: 54.6 μTs/m ; b: 84.5 μTs/m; c: 130.0)μ Ts/m. It is obvious that the profile of the double sample becomes wider when the gradients increase. This is directly due to the bigger frequency difference between the rotating nuclear spins in the neighbouring voxels of the voxel strip.

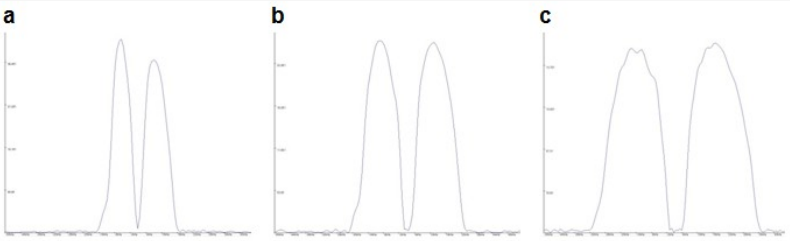

Fig. 19: Ideal profile of a double sample of oil and water with an increasing gradient strength. Because of the special sample alignment, the gradient is applied in the z-direction (a: 54.6 μTs/m ; b: 84.5 μTs/m; c: 130.0 μTs/m). The profile becomes wider when the gradient strength increases.

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A: 1D spatial encoding by frequency encoding (5/8)

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Record an ideal profile of the double sample, i.e. display the signals of oil and water in a spatially resolved manner so that both signals are visible in a separate manner. Observe this profile at different repetition times.

In accordance with Figs. 18 a-c, it is clear that the gradient field must be applied in the z-direction in order to make the individual samples separately visible (ideal profile) based on the special alignment of the double sample. Now, the influence of the repetition time of this profile is to be studied.
For all of the measurements in Figs. 18 and 19 a-c, the repetition time was set to 5 seconds. This means that the repetition time is longer than the relaxation time of water and even longer than the relaxation time of oil (see "Relaxation times in nuclear magnetic resonance"). With this long repetition time, it is not possible to clearly identify the individual samples of the double sample.

A: 1D spatial encoding by frequency encoding (6/8)

PHYWE

Figs. 20 a-c show the profile of the double sample with a gradient applied in the z-direction (72.8 μTs/m)and for three different repetition times (a: 5 s, b: 1.5 s, c: 0.5 s). In the case of shorter repetition times, it can now be clearly seen which sample has the shorter relaxation time and which sample the longer relaxation time (see Figs. 20 b and c), i.e. which of the profiles can be assigned to the oil sample (short relaxation) and which to the water sample (long relaxation). In the case of short repetition times, the magnetisation vector of the water sample cannot completely reach its state of equilibrium, i.e. the parallel alignment with regard to the external static magnetic field $\vec{B_0}$. As a result, the signal of the water sample becomes clearly weaker in the subsequent measurement (see Figs. 20 b and c).

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A: 1D spatial encoding by frequency encoding (7/8)

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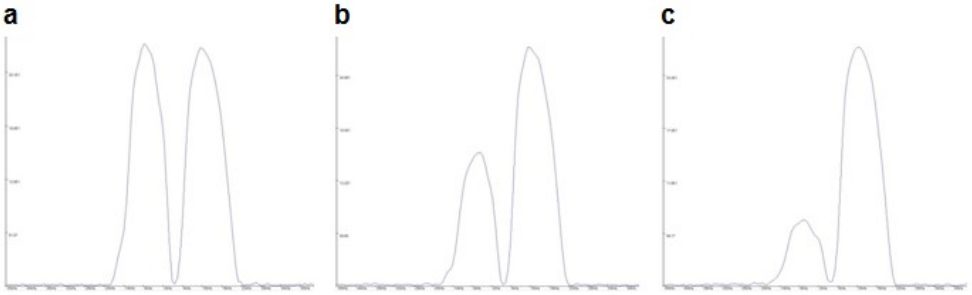

Fig. 20: Ideal profile of a double sample of oil and water measured with different repetition times. A gradient of the strength 72.8 μTs/m is applied in the z-direction. In the case of a very long repetition time of 5 s (a) that exceeds the relaxation time of oil and in particular that of water, the two samples cannot be distinguished, i.e. both samples have approximately the same signal strength. In the case of shorter repetition times, e.g. 1.5 s (b) or 0.5 s (c), both samples can be identified very well, since these short repetition times are not sufficient for bringing the magnetisation vector of the water sample back to its state of equilibrium. Consequently, the signal of the water sample becomes considerably weaker in the subsequent measurement.

A: 1D spatial encoding by frequency encoding (8/8)

PHYWE

Study the behaviour of an ideal profile of the double sample when this double sample is rotated in the sample chamber (90°, 180°).

Parts 1 and 2 have clearly shown what happens when the double sample is rotated in the sample chamber. If the sample is parallel to the rear edge of the magnet as it is with the case in part 1 and 2, the z-gradient supplies an ideal profile of both samples in which the individual samples can be clearly distinguished. If the sample is now rotated by 90° in the MR sample chamber, the z-gradient is suddenly perpendicular to the parting line between the oil and water samples and the individual samples that could be clearly distinguished beforehand are now represented as a "joint representation" of the double sample. The gradient in the z-direction could be replaced with a gradient in the x-direction and the individual samples could be distinguished again. If the double sample is rotated by 180° with regard to its initial position, the individual profiles of the double sample can also be distinguished of course. In this case, the individual profiles of the oil and water samples are simply exchanged.

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B: Visualisation of a spatiotemporal T_1 profile (1/4)

PHYWE

Once again, record an ideal profile of the double sample, i.e. display the signals of oil and water in a spatially resolved manner (see the settings of part A). Now, study the influence of the time step and number of points on the spatiotemporal visualisation of the T_1 profile in the encoding direction.

An ideal profile of the double sample (positioned parallel to the rear edge of the magnet housing) can be produced with a special setting of the repetition time (e.g. 1 s) and a gradient in the z-direction (e.g. 72.8 μ Ts/m).The oil and water samples can be clearly identified (compare the theory). Now, this profile is to be provided with a T_1-weighting and several measurements are to be performed in order to produce a spatiotemporal T_1-weighted profile. For this purpose, the sequence of three individual pulses (90°, 90°, 180°) that is described in the theory part is used, with the two 90° pulses being separated by a certain interval $\tau_{\Delta 90}$. By way of an adaptation of this time interval $\tau_{\Delta 90}$., a profile of a double sample with T_1-weighting can be produced, since after the second 90° pulse the signal is once again proportional to the restored longitudinal magnetisation. If $\tau_{\Delta 90}$.is varied in consecutive measurements, the result is a spatiotemporal T_1-weighted profile of the double sample.

B: Visualisation of a spatiotemporal T_1 profile (2/4)

PHYWE

In this part of the experiment, the time step and number of points are a measure of, $\tau_{\Delta 90}$. since the time step simply describes the incremental increase of the pulse interval between the two 90° pulses in consecutive measurements and the number of points stands for the number of measurements. A multiplication of the time step and the number of points results in the duration until the point when the spatiotemporal T_1-weighted profile of the double sample is to be recorded. If this duration is much shorter than the T_1 relaxation time of water, for example, but clearly longer than the T_1 relaxation time of oil, the maximum signal strength of oil in the spatiotemporal profile is already reached by the end of the measurement, whereas the signal strength of water is still comparably low. In order to obtain a complete spatiotemporal T_1-weighted profile of the double sample of oil and water, the time step and number of points should be set so that the maximum interval between the two 90° pulses exceeds the T_1-relaxation time of water at the end of the total measurement.

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B: Visualisation of a spatiotemporal T_1 profile (3/4)

PHYWE

Record an ideal spatiotemporal T_1 profile in the encoding direction and then comment on your result. An ideal profile is characterised by the fact that the oil and water samples can be clearly distinguished based on their different relaxation times. Identify the oil and water samples.

Fig. 21 shows an ideal spatiotemporal T_1 profile of the double sample in the encoding direction. The repetition time was set to 1 s and the gradient in the z-direction to 72.8 μTs/m. The time step (200 ms) and the number of points (20) have been selected so that the multiplication of both parameters exceeds the longer T_2 relaxation time of water (approx. 3 s). This means that at the end of the total measurement, the time interval between the two 90° pulses is 4 s. The individual samples can be clearly identified in the spatiotemporal profile of the double sample. The nuclear spins of the hydrogen protons in oil relax very quickly back to their ground state, whereas the nuclear spins of the hydrogen protons in water relax only comparably slowly.

B: Visualisation of a spatiotemporal T_1 profile (4/4)

PHYWE

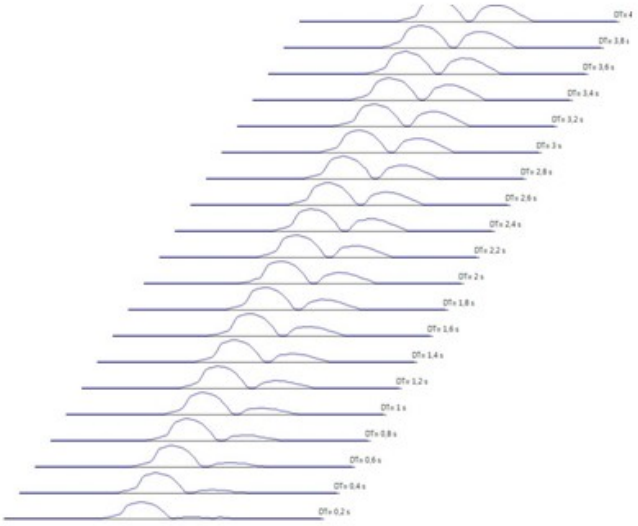

Fig. 21: Spatiotemporal T_1 profile of the double sample of oil and water in the encoding direction. The strength of the gradient in the z-direction is 72.8 μTs/m and the repetition time between the individual measurements is 1 s. The time step (200 ms) and the number of points (20) have been selected so that the multiplication of both parameters exceeds the relaxation time of water (approx. 3 s). The individual samples can be clearly identified in the spatiotemporal T_1 profile. On the right, there is the sample with the long T_1 relaxation time, i.e. water, and on the left, there is the sample with the short T_1 relaxation time, i.e. oil.

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C: Visualisation of a spatiotemporal T_2 profile (1/4)

PHYWE

Once again, record an ideal profile of the double sample, i.e. display the signals of oil and water in a spatially resolved manner (see the settings of part A). Now, study the influence of the time step and number of points on the spatiotemporal visualisation of the T_2-profile in the encoding direction.

A profile of the double sample (positioned parallel to the rear edge of the magnet housing) can be produced with a special setting of the repetition time (e.g. 2 s) and a gradient in the z-direction (e.g. 72.8 μ Ts/m). The oil and water samples can be clearly identified (compare the theory).
Now, this profile is to be provided with a T_2-weighting and several measurements are to be performed in order to produce a spatiotemporal T_2-weighted profile. For this purpose, the sequence of two individual pulses (90°, 180°) that is described in the theory part is used, with the pulses being separated by a certain interval $\tau_{\Delta 180}$ (echo time). By way of an adaptation of this time interval $\tau_{\Delta 180}$,a profile of a double sample with T_2-weighting can be produced, since after the 180° pulse (spin echo) the signal is proportional to the existing transverse magnetisation. If $\tau_{\Delta 180}$ (echo time) is varied in consecutive measurements, the result is a spatiotemporal T_2-weighted profile of the double sample.

C: Visualisation of a spatiotemporal T_2 profile (2/4)

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In this part of the experiment, the time step and number of points are a measure of $\tau_{\Delta 180}$,since the time step simply describes the incremental increase of the pulse interval between the 90° pulse and the 180° pulse in consecutive measurements and the number of points stands for the number of measurements. A multiplication of the time step and number of points results in the duration until which the spatiotemporal T_2-weighted profile of the double sample is to be recorded. We could now assume, in accordance with part B, that we would obtain a complete and real spatiotemporal and T_2-weighted profile of the oil and water double sample if the time step and number of points are set so that the maximum interval between the two pulses (90°, 180°, echo time) exceeds the T_2 relaxation time of water at the end of the total measurement. However, this is not the case. The real T_2 profile cannot be produced. If the second 180° pulse is delayed too much (e.g. up to the T_2 relaxation time of water), the signal dephases to such an extent that rephasing via the spin echo does not occur. The signal gets lost. This means that the T_2- profile gets lost much more quickly via the sequence that is described in the theory part than it could be expected based on the actual theory, and that the spatiotemporal T_2-profile is rather a T_2^{α} profile with $T_2^{\alpha} << T_2$. However, even in the case of a very quick loss the oil and water samples can be clearly distinguished.

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C: Visualisation of a spatiotemporal T_2 profile (3/4)

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Record an ideal spatiotemporal T_2 profile in the encoding direction and then comment on your result. An ideal profile is characterised by the fact that the oil and water samples can be clearly distinguished based on their different relaxation times. Identify the oil and water samples.

Fig. 22 shows a spatiotemporal T_2 profile of the double sample in the encoding direction. The repetition time was set to 2 s and the gradient in the z-direction to 72.8 μTs/m. The time step was set to 8 ms and the number of points to 20. With these parameter settings, a maximum echo time of 160 ms is generated. This maximum echo time is considerably shorter than the T_2 relaxation time of water. However, it can be seen that, due to the special sequence, the signal of the double sample has already disappeared after this time, which is due to the artificial dephasing of the nuclear spins in the time between the 90° pulse and the 180° pulse. Although the resulting spatiotemporal profile is not a real T_2 profile of the double sample and instead rather a T_2^{α} profile ($T_2^{\alpha} << T_2$), the individual samples can be clearly identified. The nuclear spins of the hydrogen protons in oil relax very quickly back to their ground state, whereas the nuclear spins of the hydrogen protons in water relax rather slowly. This means that the absolute transverse magnetisation that is measured by way of the spin echo decreases clearly more quickly in the case of oil than in the case of water.

C: Visualisation of a spatiotemporal T_2 profile (4/4)

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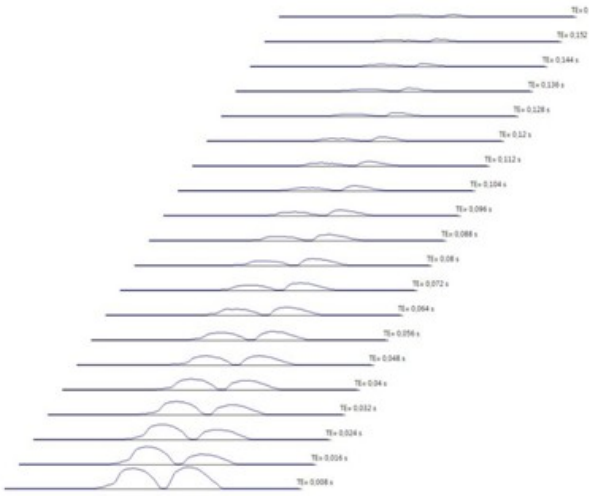

Fig. 22: Spatiotemporal T_2 profile of the double sample of oil and water in the encoding direction. The strength of the gradient in the z-direction is 72.8 μTs/m and the repetition time between the individual measurements is 2 s. The time step was set to 8 ms and the number of points to 20. With these parameter settings, a maximum echo time of 160 ms is generated between the pulses. Although the resulting spatiotemporal T_2 profile does not provide a real image of the T_2 relaxation times of oil and water (compare the artificial dephasing due to the special sequence), the individual samples can be clearly identified in the spatiotemporal T_2 profile. On the right, there is the sample with the long T_2 relaxation time, i.e. water, and on the left, there is the sample with the short T_2 relaxation time, i.e. oil.

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