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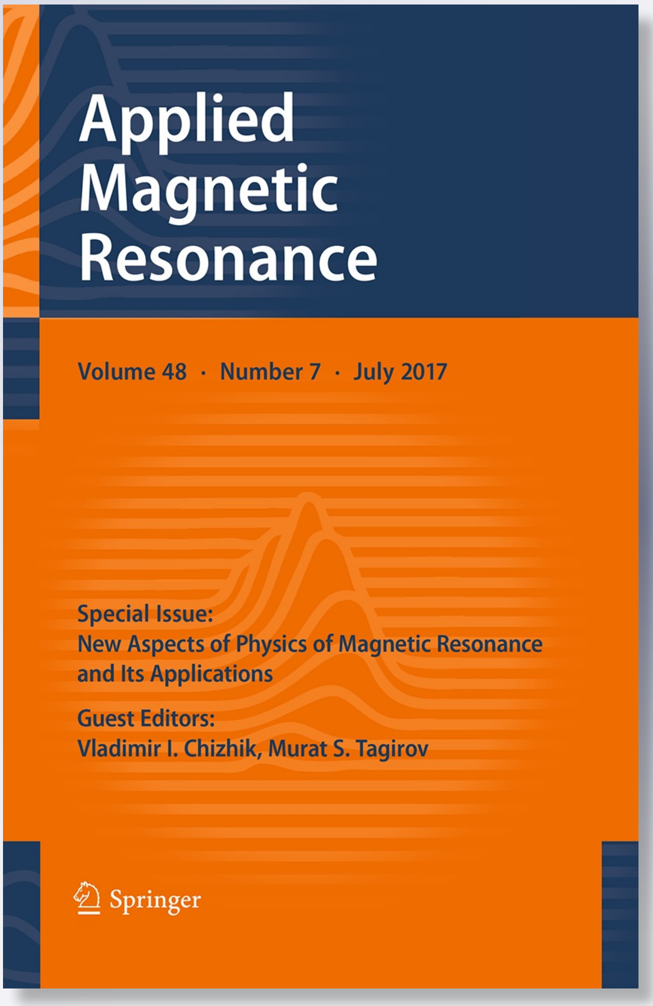

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ORIGINAL PAPER

Diffusion-Weighted Magnetic Resonance Imaging in an Ultra-Low Magnetic Field

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Abstract Magnetic resonance imaging is well known as a highly effective technique of medical visualization. One of its relatively new approaches is diffusion imaging. As a rule, the majority of magnetic resonance investigations in biology and medicine tends to be carried out in high magnetic fields (1.5 T and higher), but there are some advantages of the same experiments in low magnetic fields. It can be strongly useful, for example, for designing and testing new pulse sequences, training operators of magnetic resonance imagers, making new phantoms (model objects). In this study, diffusion-weighted imaging experiments in a low magnetic field 7 mT are performed in the first time. Nevertheless, this field is about two orders of magnitude bigger than an extremely low Earth field, and so concomitant gradients and polarization problems do not arise. In particular, diffusion weighted images of combined model samples (phantoms) are presented.

1 Introduction

By the present day, magnetic resonance imaging (MRI) has become one of the most powerful tools for medical visualization [1]. The technique is still being developed, and new branches are becoming more and more useful for medical, scientific and technical purposes. One of them, diffusion imaging, is the focus of our research.

Diffusion MRI includes two main areas: diffusion weighted imaging (DWI) and diffusion tensor imaging (DTI) containing NMR tractography. DWI is a technique which allows one to construct an MR image in which the image brightness of different tissues is determined mainly by the parameter of interest, the diffusion coefficient [2] in this case.

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The other technique, DTI [3, 4], allows us to quantitatively characterize a variation in diffusion which occurs for different spatial directions in an anisotropic media. One of the most important applications of this method is reconstruction of nerve fiber tracts in a human brain.

As the majority of other MRI techniques, DWI is based on the facts that there is a strong nuclear magnetic resonance (NMR) signal from protons of water molecules and that water is the main component of biological tissues. These facts in combination with the application of a strong magnetic field provide the rationale behind these methods effective use and applicability. Most of diffusion MRI experiments have been performed in a high magnetic field (1.5 T and higher) [4]. However, this trend is not always appropriate and has sometimes a commercial reason [1]. Some examinations can be carried out in a lower field [5]. Herewith, a better absolute homogeneity is attained, and this fact allows us to use smaller encoding field gradients. That relaxes a discomfort which is due to noise of gradient commutation and it slacks restrictions for patients with implants. There is a diminution of specific high-frequency radiation absorption rate (SAR) too. Besides, in a low magnetic field, for example, equipment mobility and a wide range of possible equipment and software modifications can be easily carried out. The low magnetic field techniques can also be used in order to decrease the cost of testing new pulse sequences, training operators of MRI machine and working with phantoms. It is particularly useful for the diffusion MRI techniques [6–9] because the diffusion coefficient does not depend on magnetic field strength. The interest to NMR in a low field is supported by development of techniques of NMR signal amplification: dynamic nuclear polarization [10], superconducting quantum interference device (SQUID) [11], hyperpolarization [12]. A specific interest to the low-field NMR and MRI techniques is stimulated although by possibility to use the technique for detection of energetic and illicit substances [13–19].

In this paper, we present the results of continuation and extension of our preliminary research on DWI [20]. This experiment was performed using a magnetic resonance imager with the magnetic field of 7 mT and relative homogeneity 10^{-5}, designed and manufactured in the section of quantum magnetic phenomena of Saint Petersburg State University. The field of 7 mT is known as "ultra-low", for example, its value is less than that given in the paper [9] by the order of magnitude. But it is bigger than the Earth field by the two orders of magnitude, so we have escaped such troubles as concomitant gradients and a necessity of polarization [6].

2 Measurements of Diffusion Coefficients

2.1 Method

The objective of our research was to prove that self-diffusion coefficient measurements can be correctly executed with the described above NMR-imager and to find out the best parameters of pulse sequences for further experiments. In order to achieve this, we have used a basic radiofrequency echo pulse sequence of 90°- and 180°-pulses separated by interval τ (the Hahn echo) with the addition of two strong magnetic field gradient pulses, also called "diffusion gradients" [21] (Fig. 1).

The diffusion gradients prevent an echo signal of moving spins from phasing-in. Thereby, the diffusion leads to attenuation of the echo signal. Measuring dependence of signal attenuation on pulse sequence parameters, one can calculate a diffusion coefficient D of an investigated object in concordance with the formula given below:

$$D = \frac{-\ln\left(s/s_0 \right)}{b}.$$

Here, s and s_0 are the signal intensities with and without diffusion gradients and the denominator, called the b-value, is [3].

$$b = (\gamma \delta G)^2 \left(\Delta - \delta/3 \right).$$

In this expression, γ is a gyromagnetic ratio, δ is a diffusion gradient duration, Δ is an interval between starts of the gradient pulses as it is indicated in Fig. 1, G is the gradient strength. The pulse sequence shown in Fig. 1 was created in the LabVIEW environment.

3 Results

Diffusion coefficients for a number of samples (water, ethanol, sunflower oil) have been measured at the temperature of 28 °C: $D_{\text{water}} = (2.7 \pm 0.1) \times 10^{-9} \frac{m^2}{s}$, $D_{\text{ethanol}} = (1.3 \pm 0.1) \times 10^{-9} \frac{m^2}{s}$ (the measured values agree with the tabular values), $D_{\text{sunfloweroil}} = (5.0 \pm 0.2) \times 10^{-10} \frac{m^2}{s}$. The purpose of these

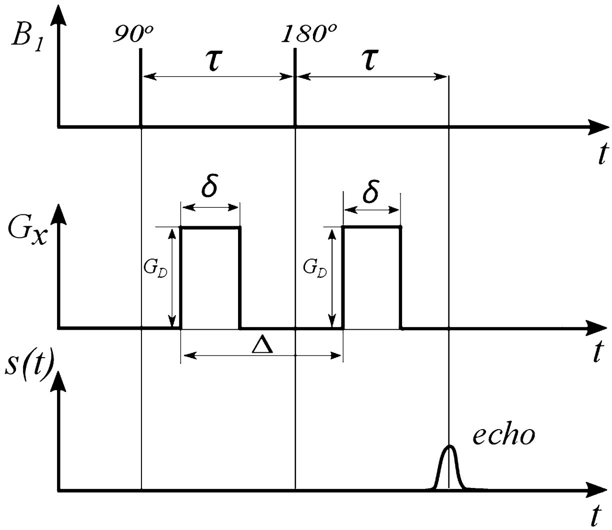

Fig. 1 Pulse sequence for diffusion coefficient measurements. B_1, G_x, $s(t)$ are a radiofrequency field amplitude, a magnetic field gradient and NMR signal, respectively, and G is the diffusion gradient magnitude. Meaning of τ, Δ and δ is clear from the diagram

measurements was to choose pairs of samples for our qualitative DWI experiment. The criterion was a significant difference in self-diffusion coefficients of the two samples. Clearly, the diffusion coefficients in pairs water–oil and oil–ethanol differed distinctly.

Diffusion coefficients for a fibrous plant sansevieria along and perpendicularly to the fibers also have been measured. The results are:

$$D_{\text{sansevieria\_lengthwise}} = (2.0 \pm 0.2) \times 10^{-9}\,\frac{m^2}{s}, D_{\text{sansevieria\_crosswise}} = (1.6 \pm 0.1) \times 10^{-9}\,\frac{m^2}{s}.$$

Therefore, we have demonstrated the ability to measure correctly self-diffusion coefficients for the samples with our ultra-low MR-imager and the ability to distinguish diffusion coefficients for different spatial directions in fibrous plants, i.e., to detect diffusion anisotropy.

4 Diffusion Weighted Imaging

As it was mentioned earlier, DWI is a technique which is based on different translational molecular mobility of different areas in investigated samples. The result of this method usage is a contrast enhancement in the image in dependence on the self-diffusion coefficient.

Contrasted images can be divided into two groups. The first is parametric images, i.e., the images of spatial distribution of the parameter, in which brightness is proportional to the value of a displayed parameter. And the second is parameter-weighted images, in which brightness of the image is determined not only by the parameter of interest, but also by other factors affecting the intensity of the NMR signal.

Diffusion weighting along with T_1-, T_2-weighting is one of the most well-known MRI techniques. The method has a great impact on medical visualisation and has been successfully implemented in clinical cases for a long time. There is a great interest for medicine in those cases when a DW image of a normal tissue differs from a DW-image of a pathologically changed one.

Although the technique has its own significant value, here we also use it to demonstrate our ability to perform multi-pixel diffusion experiments which is one of the main steps of a DTI experiment.

4.1 Method

The method of obtaining DWI represents a combination of methods for measuring diffusion coefficients and methods of spatial encoding in MRI.

The Larmor theorem can be used in order to obtain information about a position of a voxel (volume element): $\omega = \gamma B_z$, here ω—nuclear magnetic resonance frequency, γ—gyromagnetic ratio, B_z—magnetic field in which the nuclear magnetic resonance signal is observed. To "mark" a voxel coordinate with frequency one can use a gradient magnetic field with the gradient pulse applied in a direction of interest (G_x, G_y, G_z).

In the research, the pulse sequence based on a spin-echo signal formation in the presence of balanced diffusion-encoding magnetic field gradients has been used (Fig. 2). The combination of radio-frequency (RF) pulses with the amplitude B_1 (90° RF-pulse and 180° RF-pulse) and magnetic field gradient pulses (frequency-encoding gradients G_{read}, a phase-encoding gradient G_{ph}, diffusion gradients G_{D}) allows us to obtain a two-dimensional image contrasted with a diffusion coefficient after twofold Fourier transform of the echo-signal $s(t)$.

In the DW images more intense signal is observed in areas with lower molecule mobility, and weaker signal is observed in areas with higher mobility. To employ the imaging pulse sequence special programs were created in the LabVIEW environment.

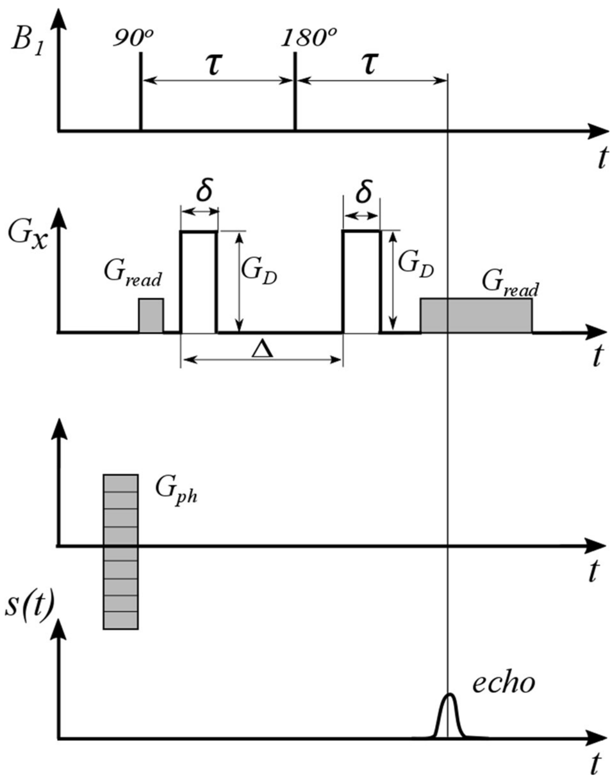

Fig. 2 Pulse sequence to obtain DWI: G_D, G_{read} and G_{ph} are diffusion, frequency-coding and phase-coding gradients, the rest *symbols* are the same as in Fig. 1

4.2 DWI Visualization

For this purpose, we obtained DW images of two combined samples: water–sunflower oil (Fig. 3) and sunflower oil–ethanol (Fig. 4). The series of images obtained from experiments with gradual increase of diffusion weighting were carried out for these combined samples. The amplitude of the diffusion gradient G_D has been changed from 0 to 1.36×10^{-2} T/m. For diffusion-weighted images an extent of sensitivity to diffusion is determined by the b-value (see Sect. 2.1).

Experiments were conducted with the following parameters: interval between RF pulses TE/2 = τ = 100 ms, repetition time TR = 800 ms, duration of 90° RF-pulse—80 μs, each measurement was averaged over four accumulations, interval between the gradient pulses Δ = 110 ms, gradient pulse duration δ = 30 ms. To obtain images without diffusion weighting gradient pulses were turned off.

In the first experiment, the ampoule with a diameter of 8 mm filled with water was placed in the cylindrical container with a diameter of 25 mm filled with oil. For the second combined sample, two cylindrical containers with diameters of 20 mm and 25 mm were placed one in the other (oil was inside, ethanol-outside).

For data processing, digital filtration and construction of images, a special program was created in the Mathcad environment.

Having analyzed the images of the *water–oil* combined sample (Fig. 3), it has become obvious that with the increase of the diffusion gradients the signal from water significantly decreases. This result is consistent with the fact that the water self-diffusion coefficient considerably exceeds the oil self-diffusion coefficient.

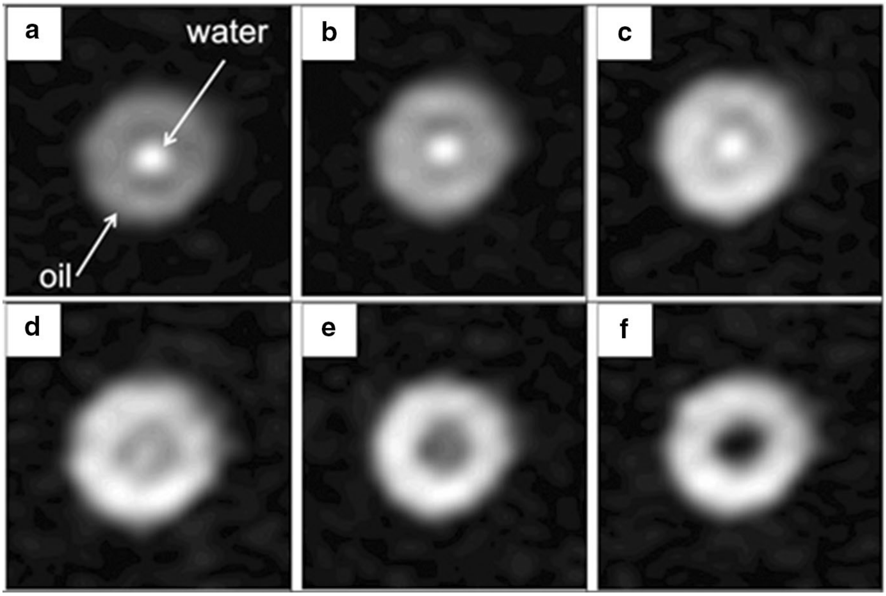

Fig. 3 Visualization of the combined sample water–oil: **a** b = 0 s/cm^2, **b** b = 10 s/cm^2, **c** b = 330 s/cm^2, **d** b = 750 s/cm^2, **e** b = 1330 s/cm^2, **f** b = 2080 s/cm^2

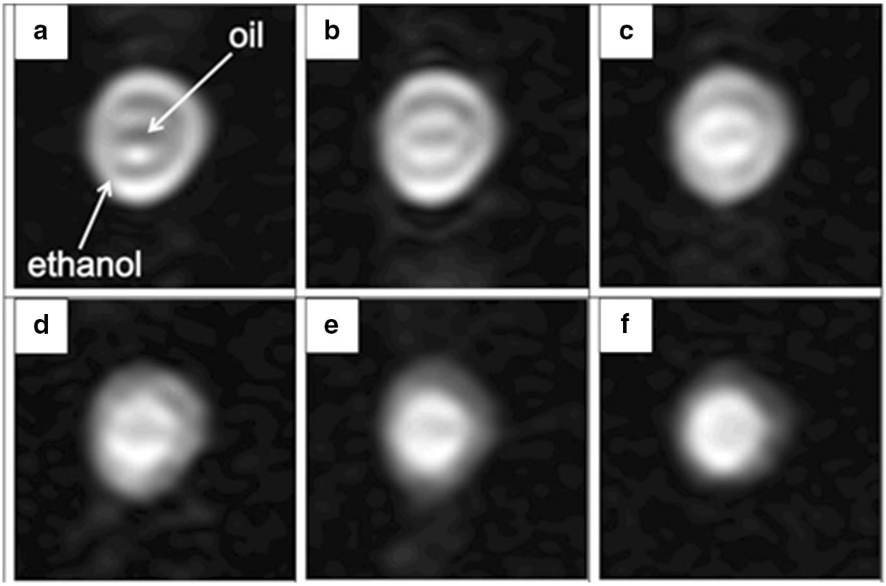

Fig. 4 Visualization of the combined sample oil-ethanol: **a** $b = 0$ s/cm^2, **b** $b = 330$ s/cm^2, **c** $b = 750$ s/cm^2, **d** $b = 1330$ s/cm^2, **e** $b = 2080$ s/cm^2, **f** $b = 3000$ s/cm^2

A similar situation is given in Fig. 4 (a combined sample *oil–ethanol*), but there are some differences. The substance with smaller self-diffusion coefficient (oil) is placed in the center of the model object. With the increase of diffusion gradients the signal from ethanol (which is located in the external part of the compound) gradually completely vanishes.

5 Conclusion

To sum it all up, we have demonstrated the ability to do diffusion weighting experiments in an ultra-low magnetic field of 7 mT in spite of considerable lack of sensibility. Herewith, it is sufficient to use a moderate strength of gradient field that has allowed us to avoid difficulties with concomitant gradients that are inherent to extremely weak magnetic field.

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