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IMAGING OF MAGNETIC FIELD DISTRIBUTION USING THIN-LAYER MAGNETIC RESONANCE METHOD

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Imaging of magnetic field distribution using thin-layer magnetic resonance methods on physical and biological samples has been performed. The resulting image represents the magnetic susceptibility distribution in the sample. An NMR imaging method susceptible to the inhomogeneity of magnetic field Gradient Echo was used. Since the investigated physical or biological samples did not generate any NMR signal, a homogeneous phantom was used - a container filled with water - as a reference medium. An image acquired by this method is actually a projection of the sample properties onto the homogeneous phantom. The method could be applied in nanotechnology, microelectronics and especially in the biological and medical sciences e.g. for drugs delivery monitoring.

Keywords: magnetic field distribution, magnetic liquids, magnetic resonance imaging, magnetic susceptibility

1 INTRODUCTION

Measurement and imaging of ferromagnetic or paramagnetic objects that do not contain any water molecules and do not produce any NMR signal is not possible using standard magnetic resonance imaging (MRI) methods. Inserting such an object into a stationary homogeneous magnetic field results in field deformation proportional to the susceptibility of the sample. If the space in the vicinity of the sample is filled with water, we are able to image this sample as a projection. In the case of a homogeneous phantom (e.g. a slender rectangular container filled with water), the acquired image represents a modulation of the local magnetic field representing the magnetic susceptibility distribution in the sample. Using an appropriate mathematical model it is possible to calculate this susceptibility exactly.

Using pulsed gradient spin echo NMR sequence, in vitro micro images of a sample, a solution of polyethylene oxide in water, were presented in [1]. A method used the divergence in gradient strength that occurs in the vicinity of a thin copper current – carrying wire.

An experiment with thin electrical wire imaging using phantom in the shape of a plastic sphere filled with agarose gel was published in [2]. Images of a phantom were obtained with and without application of electric current to a straight wire.

The goal of this paper is description of the magnetic field deformation (MFD) measurement expressing the properties and structure of the thin ferromagnetic or paramagnetic sample represented by its susceptibility in the form of an image using NMR imaging methods. A carefully tailored gradient echo (GE) magnetic resonance measuring sequence - was used. This imaging sequence is sensitive to relative local inhomogeneities caused by magnetic susceptibility distribution in an object placed into the homogeneous magnetic field.

2 PURPOSES AND METHOD

2.1 Magnetic field deformation consideration

Speaking in general, paramagnetic sample inserted into the homogeneous magnetic field causes deformation of magnetic field lines.

For an experiment one needs ideally homogeneous magnetic field generated e.g. by an air-core electromagnet. Placing a ferromagnetic or paramagnetic sample with $\mu_r = const.$ into the homogeneous magnetic field B_0, the MFD appears in the objects and in the surrounding space near the object.

Magnetic field deformation can be expressed as a deflection of the magnetic field lines (φ_1, φ_2) on the border of two isotropic media (μ_1, μ_2). Magnetic induction B_s in the sample by an influence of the sample susceptibility $\chi_s > 0$ reaches the value:

$$B_s = B_0(1+\chi_s) \tag{1}$$

Formula (1) declares that the magnetic induction increases in the sample and in its nearby vicinity. It is possible to image this differential magnetic field ΔB using the proposed technique.

Magnetic field in a vicinity of two material media boundary can be expressed by the equations:

$$\oint \mathbf{B}.d\mathbf{s} = 0 \tag{2}$$

where ds – vector of elementary area.

Using this equation for magnetic field lines, the deflection of two isotropic magnetic boundaries follows as:

$$tg\varphi_1 / tg\varphi_2 = \mu_1 / \mu_2 = \mu_{r1} / \mu_{r2} \tag{3}$$

It is evident that it is possible to calculate the magnetic field deformation values for concrete samples, but the

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goal of this short contribution is only to detect and to image this deformation.

2.2 Experimental configuration

A home-made MR imager 0.1 Tesla with horizontally oriented electromagnet, planar RF coil for thin layers and original gradient coil system was used for thin layers imaging. MR imager was controlled by S.M.I.S. console. The sample was placed in a plastic bushing and inserted into a vertical rectangular holder (phantom) filled with 0.1 wt% solution of $CuSO_4$ in distilled water. The solution of $CuSO_4$ was used for shortening the repetition time TR to 200 ms (for speeding up of data collection). The basic configuration of our experiment is shown in Fig.1.

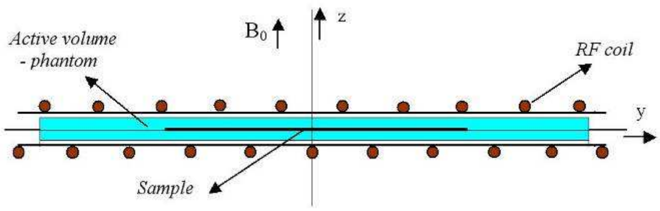

Fig. 1. Orientation of RF coil, measuring phantom and sample in the magnetic resonance imager 0.1 Tesla for thin layers imaging.

An RF transducing coil together with the sample and phantom was placed in the centre of the electromagnet perpendicular to the magnetic field (B_0) orientation. Planar RF 11 turns - coil system with interlaced wires of upper and lower part of the coil was used.

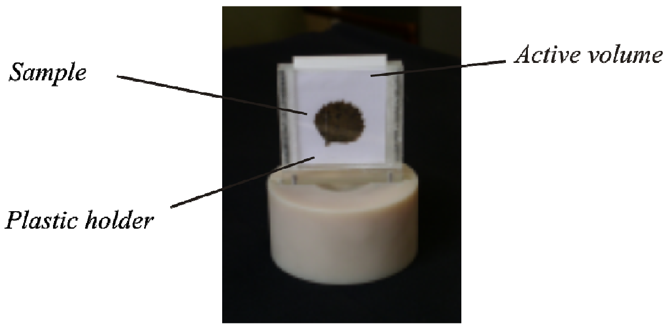

Fig. 2. The physical or biological samples were placed in a plastic holder and located into the vertical rectangular RF coil positioned into the stationary magnetic field of a magnetic resonance imager.

2.3 Modelling and testing

For testing and calibration of the method a double flat coil – meander – produced on a printed board, with a thickness of 0.5 mm, (see Fig.3) was constructed. Direct feeding currents +I and –I were selected to create a planar source of a weak magnetic field in the shape of a grid, (real dimensions 50 x 50 mm). Every individual conductor had a length *2L*. The conductor's position on the x-axis was *a* and separation of the front and rear layers (turned over 90°) was *2b*.

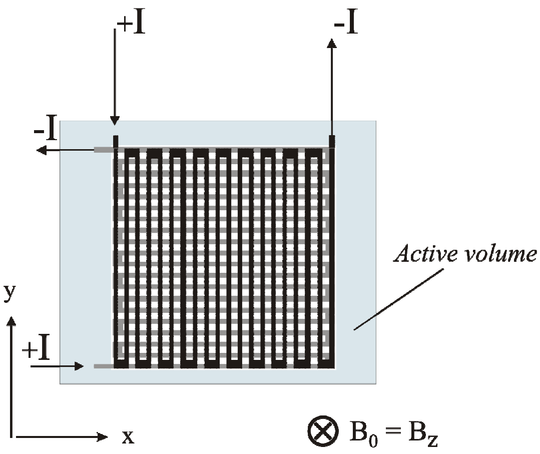

Fig. 3. Double flat coil – meander was used for testing and calibration of the method. Feeding currents +I and –I are creating a planar source of weak magnetic field in the shape of a grid.

Magnetic field generated by such a system is easy to calculate for the front layer using Biot-Savart law. Resulting formula is in the form:

$$H_z(x,y,z) = \frac{I}{\pi}\sum_{n=1}^{10}\sum_{i=1}^{2} W_{in}V_{in} \tag{4}$$

where:

$$V_{in} = \frac{b_i - y}{(a-z)^2 + (b_i - y)^2}$$

$$W_{in} = Sin\left[ArcTan\left(\frac{L-x}{a-z}\right)\right] + Sin\left[ArcTan\left(\frac{L+x}{a-z}\right)\right]$$

n – number of wires in a quadrant, first wires of the left and right quadrants are interlaced,
i - number of quadrants
V_{in}, W_{in} – parts of the formula.

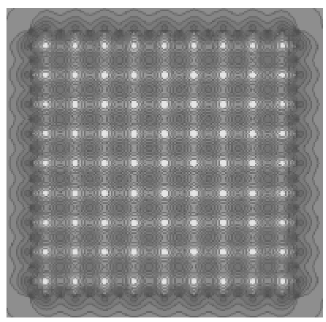

Fig. 4. Contour plot of magnetic field for y = 0 for plot points = 200, number of contours = 10, coil layer (meander): 50 x 50 mm, 19 wires in every layer.

By reciprocal exchanging the variables x and y we obtain an adequate expression for the rear layer. The resultant magnetic field is a sum of two expressions, one for the front plane (bi = b), one for the rear plane (bi = -b). The currents of the front and rear planes are oriented in opposite directions, +I and –I. Contour plot of magnetic field is depicted in Fig.4.

2.4 Imaging sequence Gradient Echo

The "Gradient Echo" NMR sequence (Fig.5) was selected for the measurement [3]. A special feature of the sequence is its sensitivity to basic magnetic field inhomogeneities interpreted in the final picture as a shadow scale.

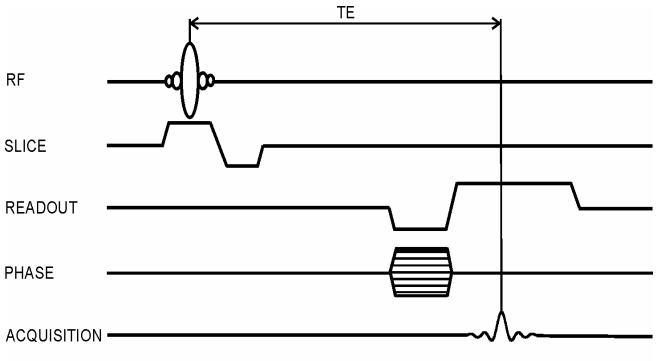

Fig.5. NMR sequence for the gradient echo signal detection. TE - echo time 32 ms.

A 5-lobe sinc pulse selectively excited the sample. The dimensions of the vessel with sample determined the slice thickness, therefore the slice gradient was switched off. Images were obtained with a field of view of 120 mm. The number of samples and the number of views determining the final resolution were 128 each and the echo time TE was 32 ms. To increase the signal-to-noise ratio of the data, the signals were averaged 16 times.

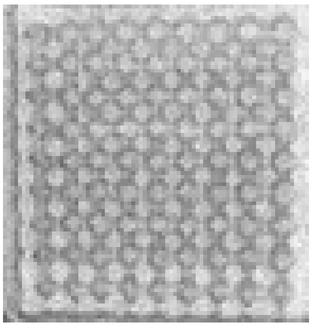

Fig. 6. Image of the MFD of a meander flat coil 50 x 50 mm, number measured samples 128 x 128, 0.1 Tesla MR imager. Actual image of this specimen was associated with 75 x 75 pixels.

In our experiment – for method calibration and testing – various feeding currents were connected to the meander coil. The maximum current was limited by the amplitude of the output NMR signal that decreased (by influence of increasing inhomogeneities) with increasing of the feeding current. For our experiment a current of 30 mA showed to be a good compromise. The first results of the magnetic field deformation imaging of the meander flat coil are depicted in Fig.6. The actual image of this sample has 75 x 75 pixels.

The model (double meander coil) served for verification of this methodology, for adjustment of basic parameters of the imaging sequence: time intervals TE (echo time, interval from RF pulse excitation to the maximum of the received NMR signal), repetition time TR (time of one measuring sequence limited by time that is necessary for spin relaxation) and number of averages. For testing, a reference environment – $CuSO_4$ solution in distilled water in connection with relaxation times variations of the measuring sequences were used.

2.5 Magnetic liquids preparation

Ferrofluids or magnetic liquids are stable colloidal suspensions composed of single-domain magnetic nanoparticles dispersed in appropriate solvents [4, 5]. In general, magnetic particles are derived from the solid solution of the spinel Mn_xFe_1-xFe_2O_4. Surfactants are added during the synthesis of magnetic liquids to surround the small particles and overcome their attractive tendencies. For aqueous-based synthesis of magnetic liquids the magnetite Fe_3O_4 and the surfactant tetramethylammonium hydroxide, $(CH_3)_4N(OH)$, were used.

In the absence of a magnetic field each particle may be considered independent and its magnetisation direction is randomly oriented, hence such system resembles a paramagnetic gas. However, when number density or magnetic moment of magnetic particle is large, one cannot neglect dipole-dipole interaction between particles. Such interaction may be manifested in the dynamical magnetic properties of magnetic liquids.

It is possible to use various methods to coat samples with a magnetic substance: soaking in a diluted magnetic liquid, using an oriented spray, steaming with magnetic steam, sputtering of the pulverized form. The goal is to assure homogeneous application of the magnetic substance onto the sample. During the soaking method magnetic liquid can be absorbed in the sample that could cause some changes of biological and physical properties. Hence, it is important to distinguish living or desiccated samples (with a minimum of original liquid). For sample preparations in our experiments, water solution of the magnetic liquid was diluted by 1:500.

2.6 Imaging of thin biological and physical layers

The first experiment on a biological sample was performed. A dry leaf was immersed into magnetic liquid solution for 24 hours. The circle-shaped pieces of blotting paper with diameter of 20 mm were used as another sample. The diluted solution of the magnetic fluid was laid on at the samples to get a scale of the magnetic susceptibility distribution in 6 grades. After drying up the samples were stuck on a sheet of paper and inserted into protective plastic bushing. The bushing with the sample was squeezed to the water phantom. The best results were achieved when the bushing was immersed into the water.

Using the imaging method Gradient Echo the slice selective gradient G_Z was omitted. The G_Y gradients ensured the phase encoding and the G_X gradients the readout of the NMR signals in the shape of the gradient echo.

The holder dimensions were 100x140x10 mm. The image resolution was 512x512 pixels and the number of averages was of 16 (Fig.7).

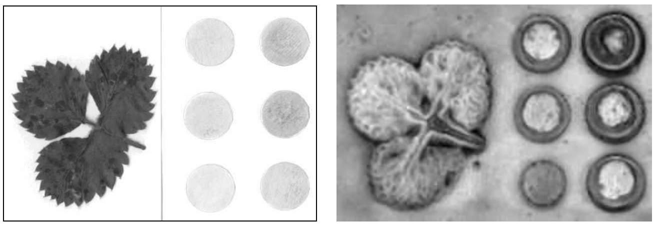

Fig.7. Left: Optical image of the sample. Right: Image of the sample as a projection into the phantom, sampling 512 x 512, resolution 100 μm.

For the next experiment on a physical sample, a banknote equipped with hidden integrated magnetic signs (for the security reason the origin is not published). The banknote was positioned in a plastic holder and placed into the vertical rectangular container (phantom) filled with 0.1 wt% solution of $CuSO_4$ in distilled water, for shortening the repetition time TR to 200 ms. The image was sampled with 512x512 samples, the number of averages was 16, the resolution 150 μm, and the static magnetic field B_0 = 0.1 Tesla, (Fig.8).

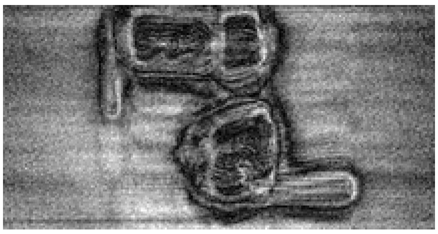

Fig. 8. Magnetic image of a banknote with visible magnetic islands, 512x512 samples.

Because of the relative high susceptibility of the magnetic signs a corona appeared around the patterns.

3 CONCLUSIONS

In the first experiments with thin biological and physical layer imaging, coated samples with a magnetic substance (magnetic liquid) and a homogeneous phantom were used. The first results showed the feasibility of the method and some of possibilities offered in this field of research.

The experiment revealed the following facts:

Linear projection of the magnetic quantities of the sample into the phantom can be performed only when the vector of the static magnetic field B_0 is perpendicular to the sample plain. Other orientations caused blurring the image edges in the z-axes direction due to magnetic field.

Reducing the thickness of the liquid in the phantom increases the sharpness and quality of the sample image but reduces the signal-to-noise ratio.

Reduction of the RF sensor (planar RF coil) and phantom vessel dimensions and using a special gradient coil system can allow the acquisition of images with high resolution (micro imaging).

From the reason of inhomogeneities of the static magnetic field it is possible to make an image correction by subtraction of the phase images of the phantom without a sample, and the phantom with a sample.

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