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MAGNETIC FIELD DISTRIBUTION MEASUREMENT OF THIN-LAYERS USING MAGNETIC RESONANCE IMAGING SEQUENCES

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Abstract: Magnetic field distribution measurement and imaging of thin layers using magnetic resonance techniques on biological samples have been performed. The resultant image represents the magnetic susceptibility distribution in the sample. An NMR imaging method susceptible to the homogeneity of magnetic field Gradient Echo was used. Since the investigated physical or biological samples did not generate any NMR signal, a homogeneous phantom (reference medium) was used - a container filled with water - as a 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.

Keywords: magnetic resonance imaging, thin-layer, magnetic susceptibility, magnetic fluids

1. INTRODUCTION

Imaging of feromagnetic or paramagnetic objects that do not incorporate any water molecules and do not generate any NMR signal is not possible using standard 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 a water containing substance, we are able to image this substance. 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 exactly calculate this susceptibility.

An experiment with thin electrical wire imaging using phantom in the shape of a plastic sphere filled with agarose gel was published in [1]. Images of a phantom were obtained with and without application of electric current to a straight wire.

Our task is to measure the magnetic field deformation (MFD) and to express 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) - NMR measuring sequence - was used.

2. PURPOSES AND METHOD

Suppose we have an ideally homogeneous magnetic field generated by an air-core electromagnet. When a ferromagnetic or paramagnetic object is placed into this homogeneous magnetic field, the magnetic induction is as follows:

$$B = \mu H = \mu_0 \mu_r H = \mu_0 H + J \tag{1}$$

where $\mu_r = \mu / \mu_0$ is relative permeability and

J - intensity of magnetisation.

From the formula (1) we can write for intensity of magnetisation:

$$J = \mu_0 \mu_r H - \mu_0 H = \mu_0 (\mu_r - 1) H = \chi H \tag{2}$$

Where $\chi = \mu_0 (\mu_r - 1)$ is an absolute susceptibility of the ferromagnetic object [2].

In general, placing a ferromagnetic or paramagnetic sample with $\mu_r = const.$ into the homogeneous magnetic field, the MFD appears in the objects and in the surrounding space near the object.

Suppose we insert a thin paramagnetic sample into a homogeneous magnetic field perpendicularly to the field lines.

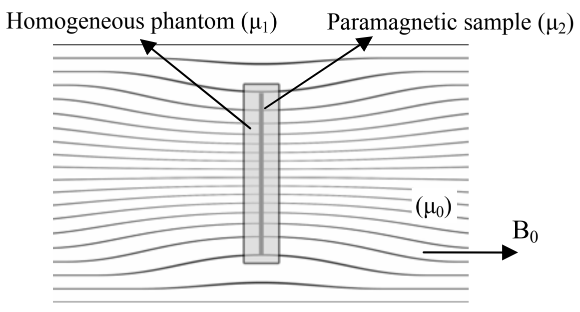

Fig. 1. Paramagnetic sample inserted into the homogeneous magnetic filed causes deformation of magnetic lines of force.

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

$$B_s = B_0 (1 + \chi_s) \tag{3}$$

This means that the magnetic induction increases in the sample and is decreasing in the sample vicinity. It is possible to image this differential magnetic field ΔB using the proposed technique.

There is no exact measurement of the magnetic susceptibility in this first step. It is only an attempt to image thin paramagnetic samples.

In general, magnetic field in a vicinity of two material media boundary can be expressed by the equations:

$$\oint \mathbf{B}.d\mathbf{s} = 0 \quad \text{and} \quad \oint \mathbf{H}.d\mathbf{s} = 0 \qquad (4)$$

where *ds* – vector of elementary area.

Using these equations for magnetic field lines, the deflection of two isotropic magnetic boundaries follows as:

$$tg\varphi_1 / tg\varphi_2 = \mu_1 / \mu_2 = \mu_{r1} / \mu_{r2} \qquad (5)$$

It is evident that it is possible to calculate the magnetic field deformation values for concrete samples, but the goal of this contribution is only to detect and to image this deformation.

2.1. *Experimental configuration*

In our experiment two MR imagers with different fields were used (0.1 and 4.7 Tesla). The sample was placed in a plastic holder (phantom) and placed into a vertical, rectangular container filled with 0.1 % 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. 2.

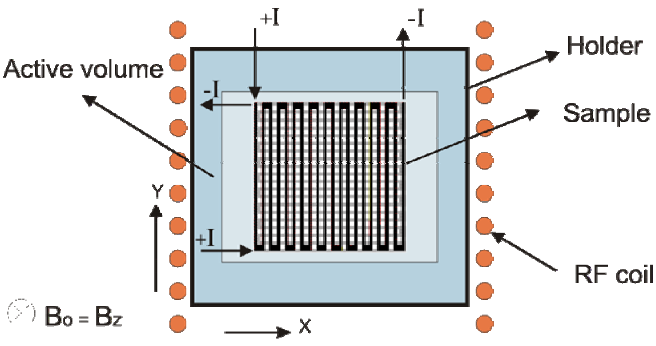

Fig. 2. Orientation of the RF coil, active measuring volume (holder – phantom) and sample in the magnetic resonance imager.

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. A solenoid detection coil was designed with the goal to optimize the RF field homogeneity.

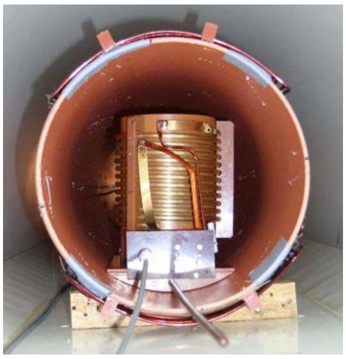

Fig. 3. RF coil located into the magnetic field of a magnetic resonance imager. A cylinder around RF coil represents a gradient coil system designed special for this experiment.

2.2. *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.2) was selected. Feeding currents +I and –I were selected to create a planar source of magnetic field with the shape of a grid. 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 90^0) was *2b.* The magnetic field generated by such a system for the front layer is expressed by [3]:

$$H_z(x, y, z) = \frac{I}{\pi} \int_{-a}^{a} W_{in} V_{in} da \qquad (6)$$

where:

$$V_{in} = \frac{b_i - y}{(a - z)^2 + (b_i - y)^2} \qquad (7)$$

$$W_{in} = Sin\left[ArcTan\left(\frac{L - x}{a - z}\right)\right] + Sin\left[ArcTan\left(\frac{L + x}{a - z}\right)\right] \qquad (8)$$

By reciprocal exchanging the variables *x* and y we obtain an adequate expression for the back layer. The resultant magnetic field is a sum of two integrals, one for the front plane (b_i = b), one for the rear plane (b_i = -b). The feeding currents of the front and rear planes are oriented in opposite directions, +I and –I. The integrals were computed as sums.

Fig 4. shows the density plot of the magnetic field for *y=0* (the middle plain between front and rear layers), using 75 points. The dimensions of the coil layer (meander) were: 50 x 50 mm, 19 wires in every layer symmetrical to the central axis.

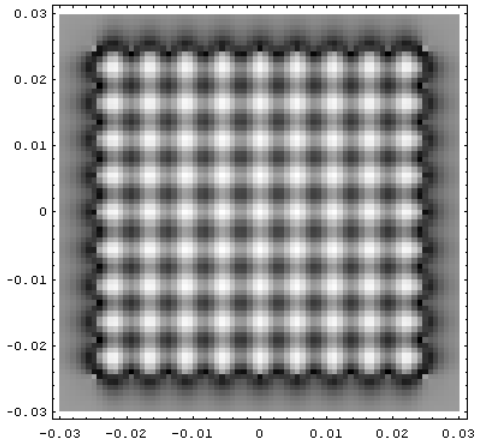

Fig. 4. Density plot of magnetic field for y = 0 for plot points = 75, coil layer (meander): 50 x 50 mm, 19 wires in every layer.

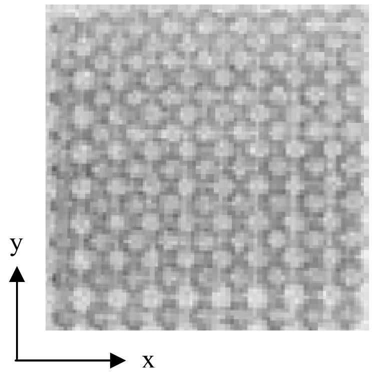

Fig. 6. Image of the MFD of a meander flat coil 50 x 50 mm, number measured samples 128 x 128. Actual image of this specimen was associated with 75 x 75 pixels.

2.3. Imaging sequence

The "Gradient Echo" NMR sequence (Fig. 5.) was selected for the measurement [4]. A special feature of the sequence is its sensitivity to basic magnetic field inhomogeneities.

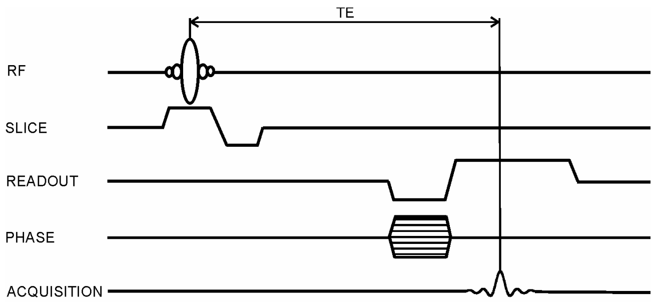

Fig.5. Simplified time diagram of the NMR sequence of the gradient echo.

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 was 32 ms. To increase the signal-to-noise ratio of the data, the signals were accumulated 16 times.

In the first step – for method calibration and testing – different 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) as a function of increasing feeding current. For our experiment a current of 30 mA showed to be a good compromise. The first results of MFD imaging of the meander flat coil are depicted in Fig.6. The actual image of this sample has 75 x 75 pixels. For comparison the same imaging structure as the model (plot points = 75) was selected.

The model (double meander coil) served for verification of this methodology, for the adjustment of basic parameters of the imaging sequence: time intervals TE and TR, number of averages, and for testing of a reference environment – $CuSO_4$ solution in distilled water in connection with relaxation times of the measuring sequence.

2.4. Using magnetic fluids

Magnetic fluids (ferrofluids) are stable colloidal suspensions composed of single-domain magnetic nanoparticles dispersed in appropriate solvents [5], [6]. 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 ferrofluids to surround the small particles and overcome their attractive tendencies. For aqueous-based synthesis of ferrofluids 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 fluids.

It is possible to use various methods to coat samples with a magnetic substance: soaking in a diluted magnetic fluid, 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 fluid 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 fluid was diluted by 1:500.

2.5. Imaging of thin layers

The first experiment on a biological sample was performed. A magnetic fluid with a dilution factor of 1:500 for a biological sample preparation was used. A dry leaf was immersed into such solution for 24 hours. After drying 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 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 60x60x10 mm. The image resolution was 256x256 or 512x512 pixels and the number of accumulations was of 16, (Fig. 7).

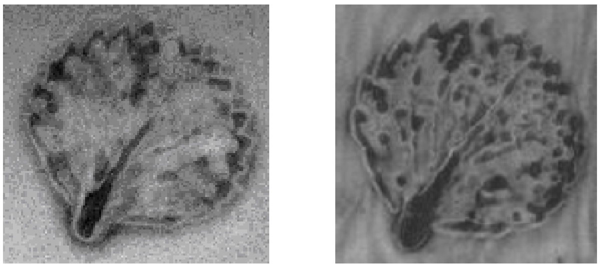

Fig.7. Sample dimension: leaf Φ = 30 mm.
Left: sampling 256 x 256, resolution 230 μm, 0.1 Tesla
Right: sampling 512 x 512, resolution 100 μm, 4.7 Tesla

For the next experiment on a physical sample, the letters "**NMR**" with the dimensions of 55x17 mm were printed using a magnetic liquid on a sheet of standard business paper. The sample was positioned in a plastic holder and placed into the vertical rectangular container (phantom) filled with 0.1 % 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 accumulations was 16, the resolution 150 μm, and the static magnetic field B_0 = 0.1 Tesla, (Fig. 8).

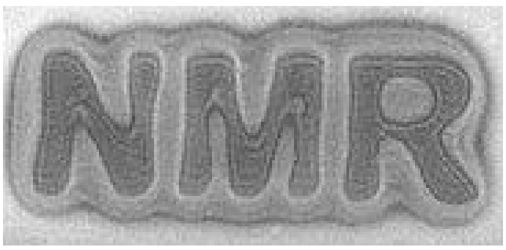

Fig. 8. Image of letters "NMR", dimension 55x17 mm, 512x512 samples

Because of the relative high susceptibility of the printed letters a corona appeared around the pattern.

3. CONCLUSION

The first experiments on thin biological and physical layer imaging were performed using a magnetic fluid and a homogeneous phantom. 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:

1. 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.

2. 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.

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

4. Image correction, from the reason of inhomogeneities of the static magnetic field is possible by subtraction of the phase images of the phantom without a sample, and the phantom with a sample. In our case, because of the relatively high susceptibility of the samples, a resultant image correction was not necessary.

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