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ELECTROMAGNETIC PHANTOM DESIGN FOR MEASUREMENT AND IMAGING QUALITY TESTING USING NMR IMAGING METHODS

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Abstract − Electromagnetic phantom design for measurement and imaging quality testing using NMR imaging has been performed. First attempts of electromagnetic phantom computation and testing on an experimental NMR 0,1 T imager were accomplished. The existing geometrical and chemical phantoms are generally used for testing of NMR imaging systems. They are simple cylindrical or rectangular objects with different dimensions and shapes with holes filled with specially prepared water solutions. In our experiments a homogeneous phantom (reference medium) was used - a container filled with water - as a standard. The resultant image represents the magnetic field distribution in the homogeneous phantom. For detection a standard gradient-echo imaging method, susceptible to magnetic field homogeneity, was used. An image acquired by this method is actually a projection of the sample properties onto the homogeneous phantom.

The goal of the paper is to map and image the magnetic field deformation using NMR imaging methods. In our experiments we are using a slender rectangular vessel with constant thickness filled with specially prepared water. A carefully tailored gradient-echo NMR measuring sequence was used.

Keywords: magnetic phantom, magnetic resonance imaging, quality testing

1. INTRODUCTION

The Nuclear Magnetic Resonance (NMR) measurement and imaging of proton density of biological structures needs to create basic physical conditions, stationary, gradient and radio-frequency (RF) magnetic fields. A biological or physical object placed into the centre of a measurement region is absorbing the energy during the RF excitation pulse generated by a coil supplied by an external source - RF transmitter. After absorption is finished, the object radiates the energy back to the space. This energy is detected by an RF receiving coil, signal is amplified, demodulated and after A/D conversion submits data. Data accumulated in the buffer of a computer are processed by Fast Fourier Transform and organized in a matrix. The matrix represents a spatial distribution of the measured parameters (proton density, relaxation times, etc.).

Imaging of 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 that is 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 sample. 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.

In this paper we propose several electromagnetic phantoms, wires configurations supplied by small stabilised DC current.

Using an appropriate mathematical model it is possible to calculate the magnetic field generated by the phantoms, but in a real NMR imaging only selected magnetic field distribution components are imaged.

First attempt of a direct measurement of the magnetic field created in living tissue by a simple wire fed by a current was reported in [1]. This report describes an experiment in which magnetic fields produced by a small current applied to the forearm of a living subject have been detected in the tissue. Initial experiments were performed in vitro using a cylindrical phantom. A current was passed along the inner of two concentric cylinders filled with conductive saline and a spin-echo NMR imaging sequence was used for signal detection. The phase-image computed from the real and imaginary images demonstrated the presence of a magnetic field in the cylinder.

A method utilizing the divergence in gradient strength that occurs in the vicinity of a thin current-carrying copper wire was introduced in [2]. Using pulsed gradient spin-echo NMR sequence and a solution of polyethylene oxide in the water, in vitro micro images of a sample were presented. The paper demonstrated the effect of molecular diffusion in the vicinity of a thin wire subjected to current pulses. This effect was demonstrated in measurements on polymer solutions and liquid crystals.

A simple experiment with thin, pulsed electrical current–carrying wire and imaging of a magnetic field using a plastic sphere filled with agarose gel as phantom was published in [3]. Images of the phantom were obtained with and without application of electric current to a straight wire using a spin-echo NMR sequence. Because the method proposed in this study does not use phase images, phase unwrapping was not required. The method is designed for the detection of electric currents in biological tissues by observations of surface potentials or surrounding magnetic fields.

Nuclear magnetic resonance imaging is an effective method for indirect mapping of magnetic field distribution. Based on these principles we propose electromagnetic phantoms as an additional tool for measurement and imaging quality testing using NMR imaging methods. First attempt of the indirect susceptibility mapping of thin-layer samples using phantoms was described in [4].

For the first attempts, for testing and calibration, a rectangular and double flat coil-meander fed by electric currents and generating a weak magnetic field were used.

Our task is to map and image the magnetic field deformation using NMR imaging methods. In our experiments we are using a slender rectangular vessel with constant thickness filled with specially prepared water. A carefully tailored gradient-echo NMR measuring sequence was used.

2. METHOD OF COMPUTATION

In the next analysis we suppose two settings:

1. All conductors are creating a rectangle parallel with y axis in the rectangular coordinate system (x, y, z).
2. The diameter of conductors and the influence of supply wires are neglected.

According to Fig. 1, we assume four conductors creating a rectangle defined by four points (P, Q, T, R) with the left - right symmetry. Conductors are fed by currents +I, -I. The static magnetic field $\mathbf{B_0}$ of the NMR imager is parallel with the z-axis.

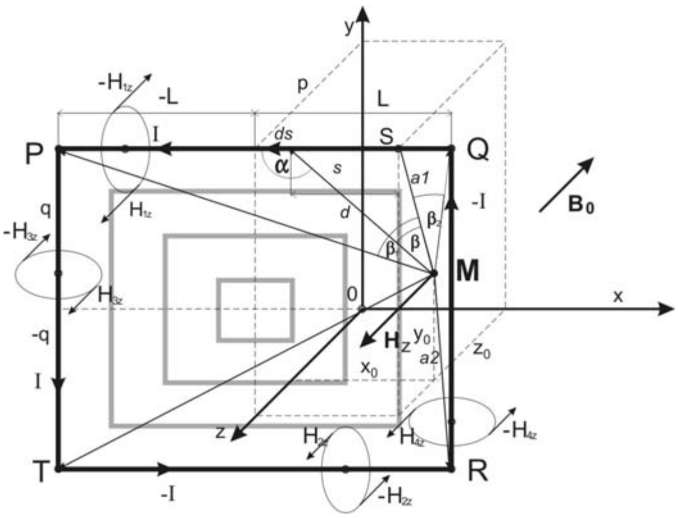

Fig. 1. Basic configuration of the four conductor rectangular coil. Three additional rectangular coils are depicted in grey colour.

Let us shortly explain the way of computation. For simplicity we consider only one horizontal conductor fed by currents +I.

Let us derive a formula for magnetic field in the point M(x,y,z). For computation of the magnetic field H_z (x,y,z) the Biot-Savart law was used [5]

$$\mathbf{H} = \frac{I}{4\pi}\oint \frac{\mathbf{ds}\times\mathbf{d}}{d^3} , \qquad (1)$$

where **d** is a distance from the point M to the conductor element ds. Considering that a_1 is the distance from the point M to the conductor PQ and β is an angle between a_1 and d, we can write the final general formula for the magnetic field differential equation as follows

$$dH = \frac{I}{4\pi}\frac{Cos\beta}{a_1}d\beta \ . \qquad (2)$$

For the distances a_1, $\overline{MP}$ and $\overline{MQ}$ we can write:

$$a_1 = \sqrt{(p-z)^2 + (q-y)^2} \qquad (3)$$

$$\overline{MP} = \sqrt{(-L-x)^2 + (q-y)^2 + (p-z)^2}$$

$$\overline{MQ} = \sqrt{(L-x)^2 + (q-y)^2 + (p-z)^2} \ .$$

After mathematical arrangements of computing of all four conductors and summing the particular field components

$$H_{z1} + H_{z2} + H_{z3} + H_{z4} = H_z \qquad (4)$$

the resultant formula has the following form

$$H_z(x,y,z) = \frac{I}{\pi}\sum_{n=1}^{4}\sum_{i=1}^{4} W_{in}V_{in} \qquad (5)$$

where

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

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

where

n – index determinating a numerical order of a wire in a quadrant,
p_n - position of a conductor on the z-axis for n = 1 to 2 both for horizontal and vertical conductors,
q_i – half length of the vertical conductors,
i - number of conductors,
L - half length of the horizontal conductors.

By reciprocally exchanging the variables x and y we obtain an adequate expressions both for horizontal and vertical conductors. The resultant magnetic field is a sum of four components according to formula (4). The current I of both horizontal and vertical conductors is in anticlockwise direction.

For the phantom practical use it is necessary to construct a coil equipped with more wires, more turns. Computation results of the rectangular coil were interpreted with the 3D-plot presentation of the magnetic field H_z (x,y,z).

For a simple example, a concentric rectangular four-coil system was designed. Following Fig. 1, next relative parameters were chosen: I = 100; p = 2; z = 1.8; and relative dimensions of the four rectangular coils: $q_1 = 1$; $L_1 = 1$; $q_2 = 2$; $L_2 = 2$; $q_3 = 3$; $L_3 = 3$; $q_4 = 4$; $L_4 = 4$.

Computation results in the form of the 3D-plot of the magnetic field H_z (x,y,z) of concentric rectangular four-coil system is depicted in Fig. 2.

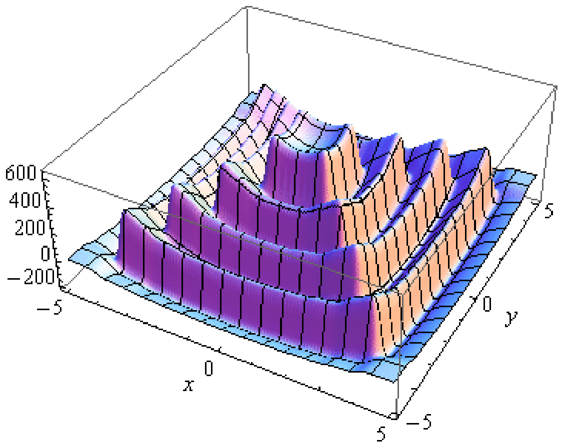

Fig. 2. 3D-plot of relative values of the magnetic field H_z (x,y,z) of the concentric rectangular four-coil system.

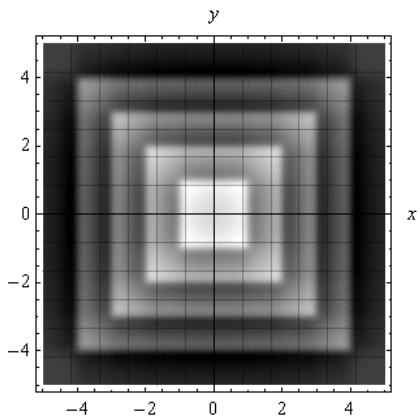

Fig. 3. Density-plot of relative values of the magnetic field H_z (x,y,z) of the concentric rectangular four-coil system.

Using this phantom for 2-D NMR imaging a density-plot presentation of the magnetic field distribution is more suitable, see Fig. 3.

3. EXPERIMENTAL RESULTS

We have used an NMR imager 0,1 T for our experiment. The phantom coil was placed in a plastic bushing and inserted into a vertical, rectangular holder filled with 0,1 wt% solution of $CuSO_4$ in distilled water. $CuSO_4$ was used to shorten the repetition time TR to 200 ms due to its reduction of H_2O T_1 (for speeding up the data collection). The general configuration is shown in Fig. 4. In our experiments we used a holder filled with doped water (inner dimensions 60 x 60 mm, constant inner thickness 10 mm).

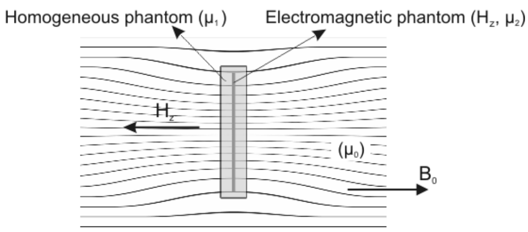

Fig. 4. Planar electromagnetic coil inserted into the homogeneous magnetic field causes deformation of the basic magnetic field B_0 lines. The coil is immersed into the holder with water.

A multi-turn, double layer planar coil was designed as a flexible phantom that fulfilled all experimental requirements for a test calibration of our experimental 0,1 T NMR imager.

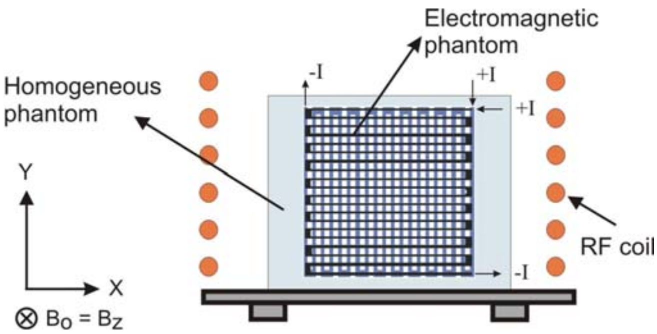

Fig. 5. Orientation of the RF coil, active measuring volume (homogeneous phantom) and sample (double flat coil) in the magnetic resonance imager.

An RF transducing coil (solenoid) together with the phantom was placed into the centre of the electromagnet, perpendicular to the magnetic field (B_0) orientation.

Using equation (5) a mathematical calculation of the magnetic field H_z (x,y,z) of the double layer planar coil was performed. The density plot of the magnetic field was calculated and depicted in Fig. 6. Feeding currents +I and –I were selected to create a planar source of a weak magnetic field in the shape of a grid. Every individual conductor had a length of *2L*.

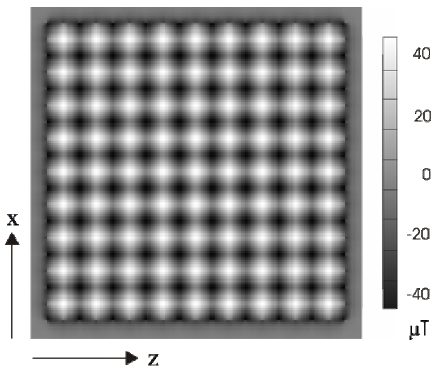

Fig. 6. Density plot of magnetic field for y = 0 for coil layer (meander): 50 x 50 mm, 19 wires in every layer, distance of layers of conductors b=1 mm.

The "Gradient-Echo" NMR sequence (Fig. 7) was selected for the measurement [6]. A special feature of the sequence is its sensitivity to basic magnetic field inhomogeneities.

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 was 128 each and the echo time TE was 32 ms. In order to increase the signal-to-noise ratio of the data, the signals were averaged 16 times. For magnetic field deformation representation a phase image was calculated using the phase unwrapping method [4].

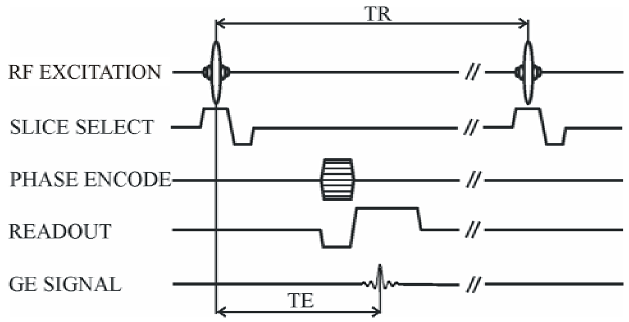

Fig. 7. Time diagram of the NMR sequence for the gradient-echo signal detection. TE - echo time of 32 ms, TR – repetition time of 200 ms.

For our experiment a current of 30 mA showed to be a good compromise. The first results of imaging of the magnetic field deformation of the meander flat coil are depicted in Fig. 8. The actual image of this sample has 64 x 64 pixels.

The phantom (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 reference environment testing – $CuSO_4$ doped water in connection with relaxation times of the measuring sequence.

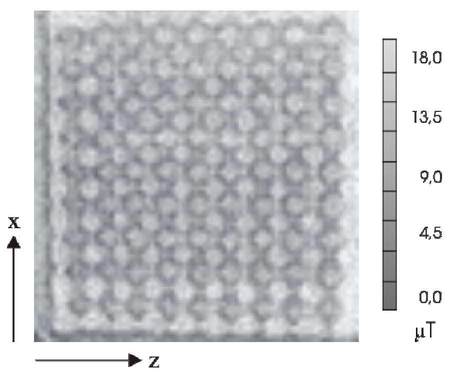

Fig. 8. Image of the magnetic field distribution of a meander flat coil (50 x 50 mm), number of measured voxels 128 x 128, actual image size was associated with 64 x 64 pixels.

The experiment revealed the following facts:

It is evident that imaging of the magnetic field of the electromagnetic phantom can be performed only if the vector of the static magnetic field $B_0=B_z$ is perpendicular to the phantom plain. Other orientations caused significant blurring of the image edges in the x- and y-axis directions due to strong basic magnetic field B_0.

4. CONCLUSION

A new method for mapping and imaging of planar electromagnetic phantoms placed into the homogenous magnetic field of an NMR imager is proposed. The method is based on a projection of magnetic field of the electromagnetic phantom into a homogeneous planar phantom (slender vessel of specifically prepared water) and subsequent NMR imaging using a gradient-echo sequence.

The method was tested using a double flat coil – meander fed by electric currents and generating a weak magnetic field.

Electromagnetic phantoms have become an additional tool for measurement and imaging quality testing using NMR imaging methods. The advantages of the electromagnetic phantoms are: universality, stability, repeatability, simple modification of basic parameters and precision. They are suitable for S/N ratio testing and very useful for imaging pulse sequences adjusting and examination.

The first results showed the feasibility of the method and some of the possibilities offered in this field of research. The electromagnetic phantoms can serve as additional quality testing tools alongside geometrical and chemical phantoms generally used in NMR imaging systems.

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