Mapping the Induction of Magnetic Field Around an Applicator for Dermato Magnetotherapy, using the Magnetic Resonance Method

[1] **Ing. Karel Bartusek, DrSc.,** [2] **Ing. Eva Gescheidtova, CSc.**

[1]Institute of Scientific Instruments,Academy of Sciences of the Czech Republic,
Královopolská 147, 612 00 Brno,
E-mail: bar@isibrno.cz
[2]Faculty of Electrical Engineering and Communication, Brno University
of Technology, Kolejni 2906/4, 612 00 Brno, Czech Republic,
e-mail: gescha@feec.vutbr.cz

Abstract. *The paper deals with measuring magnetic induction by imaging techniques based on magnetic resonance (MR) and describes experimental results of mapping the magnetic induction of a dermatological applicator for magnetotherapy performed by MR imaging techniques. The results are compared with theoretical calculation and with measurement carried out with an ordinary magnetometer. The conclusions derived from the measurement will be used in the design of a new applicator with complex electronic control of magnetic field.*

Keywords: Nuclear Magnetic Resonance, Magnetotherapy, Gradient Echo, Spin Echo

1. Introduction

To measure magnetic induction in applicators for magnetotherapy, many types of magnetometer operating on different principles can be applied. The applicators are mostly of larger dimensions and measuring is easy to realize. For magnetotherapy as applied in dermatology or cosmetics a new applicator has been developed, which consists of coils positioned in a matrix of large area and excited by suitably chosen pulses. The size of individual coils is usually comparable with the dimensions of probes used to measure magnetic induction. In these cases, measuring on the principle of magnetic resonance can be used for a precise mapping of magnetic induction around the applicator. Using this technique, an MR image weighted by the magnetic induction value can be measured with sufficient resolution in a defined layer near the applicator. Magnetic field changes due to the current flowing through the applicator can then be determined from the image.

The paper describes experimental results of mapping the magnetic induction of a dermatological applicator for magneto therapy performed by MR imaging techniques. The results are compared with theoretical calculation and with measurement carried out with an ordinary magnetometer. The conclusions derived from the measurement will be used in the design of a new applicator with complex electronic control of magnetic field.

2. MR measuring method

To map the induction of magnetic field by the magnetic resonance method the gradient echo method (GE method) can be used with advantage [1]. This is due to its sensitivity to the inhomogeneity of basic magnetic field, which deforms the image and is encoded in the image phase. A pulse sequence characterizing the measuring procedure is shown in Fig. 1.

The nuclei of the subject being measured are excited by a $\pi/2$ rf pulse of 4 ms in the presence of a delineation gradient G_S (in the direction of axis z). With a gradient amplitude $G_S = 20$ mT/m and a frequency width of rf pulse $\Delta\omega = 2700$ Hz, nuclei are excited in a layer 2 mm thick. The magnitude of the offset of rf pulse determines the position of excited layer in space. The gradient echo is produced by gradient G_R (in the direction of axis x) and occurs at time TE after the application of rf in the excitation pulse. Gradient G_P (in the direction of axis y) is responsible for the space co-ordinate y being phase-encoded in the phase of the MR signal being measured. The signal being measured can be described by the relation

$$M_T(k_x,k_y)=\iint\limits_{xy} m(x,y)\mathrm{e}^{-\mathrm{j}(k_x x+k_y y)}\,\mathrm{e}^{-\frac{TE}{T_2^*}}\,\mathrm{e}^{-\gamma\,\Delta B(x,y)TE}\,\mathrm{d}x\mathrm{d}y\,, \tag{1}$$

where

$k_\mathrm{x}=\gamma G_\mathrm{R}t'=\gamma\int_0^{t'}G_\mathrm{R}(t)\mathrm{d}t$, $k_\mathrm{y}=\gamma G_\mathrm{Pn}T_\mathrm{y}=\gamma\int_0^{t'}G_\mathrm{Pn}(t)\mathrm{d}t$	are the co-ordinates of *k*-space,
$m(x,y)$	is a function of the spin density of measured specimen in the excited plane,
$\Delta B(x,y)$	is a function of the inhomogeneity of basic magnetic field,
TE	is the time of the symmetry centre of gradient echo,
T2*	is the relaxation time when the inhomogeneity of basic field is taken into consideration.

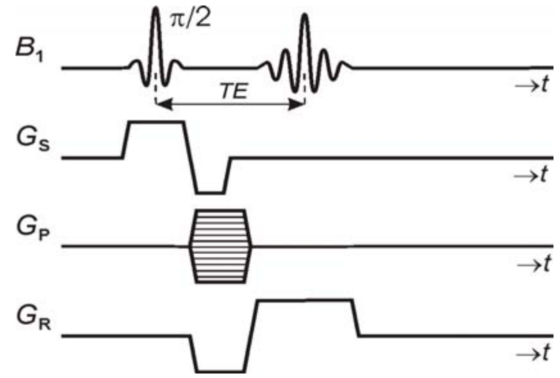


Fig. 1. Pulse sequence of gradient echo.

A 2D Fourier transform yields an image of the section through the specimen being measured, weighted by the spin density function, in the form

$$m(x,y)=\frac{1}{2\pi}\int_{k_x}\int_{k_y}M_T(k_x,k_y)\mathrm{e}^{\mathrm{j}(k_x x+k_y y)}\mathrm{e}^{-\frac{TE}{T_2^*}}\mathrm{e}^{-\mathrm{j}\gamma\,\Delta B(x,y)TE}\mathrm{d}k_x\mathrm{d}k_y\,. \tag{2}$$

The exponential in relation (2) $\mathrm{e}^{-\mathrm{j}\gamma\,\Delta B(x,\,y)\,TE}$ expresses encoding the inhomogeneity of basic field into the image phase. A change in the image phase by 2π corresponds to the change in magnetic induction

$$\Delta B_{(2\pi)}=\frac{1}{\gamma'\,TE}\,, \tag{3}$$

where

γ’	is the gyromagnetic ratio of the nucleus being measured [Hz/T].

After rewriting for an arbitrary phase change $\Delta\Phi$ the magnetic induction will be

$$\Delta B=\frac{\Delta\Phi}{\gamma'\,T_E}\,. \tag{4}$$

To map the magnetic field by the MR method it is necessary to detect the MR image in the selected plane near the coil. The phase image measured is weighted by the magnitude of the magnetic induction being measured. To determine the waveform of the z-component of magnetic induction $B_z(x,y)$ in a plane perpendicular to the coil axis it is necessary to have a thin layer of water close to the coil being measured or to immerse the coil in water and to excite the nuclei in the selected layer near the coil. The experimental set-up is shown in Fig. 2. An advantage of the mechanical delineation of the layer being measured consists in its defined dimensions and position.

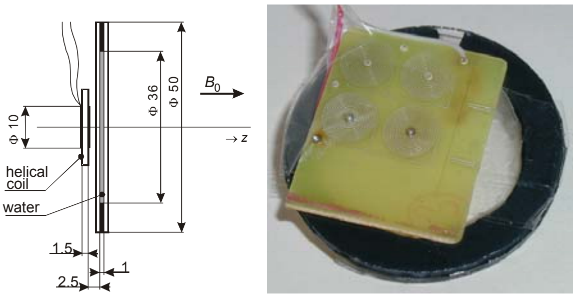


Fig. 2. Experimental set-up – the coil being measured and the mechanically delineated.

Unlike in the electronic selection of layer, large changes in the magnetic induction of the field being measured do not result in deforming the delineated layer. In the case of electronic delineation of the layer a change of 5 μT in the magnetic induction of basic field will deform the delineated layer by as much as 4 mm, which is a deformation out of all proportion. For this reason the helical coil was placed at a distance of 2.5 mm from the centre of the specimen filled with water. The specimen is made up of glass plates placed in parallel at a distance if 1 mm from each other. The gap is set by a rubber annulus and the resulting cavity is filled with water.

Fig. 3 gives the MR phase image detected by the GE method for a layer of 1 mm in thickness.

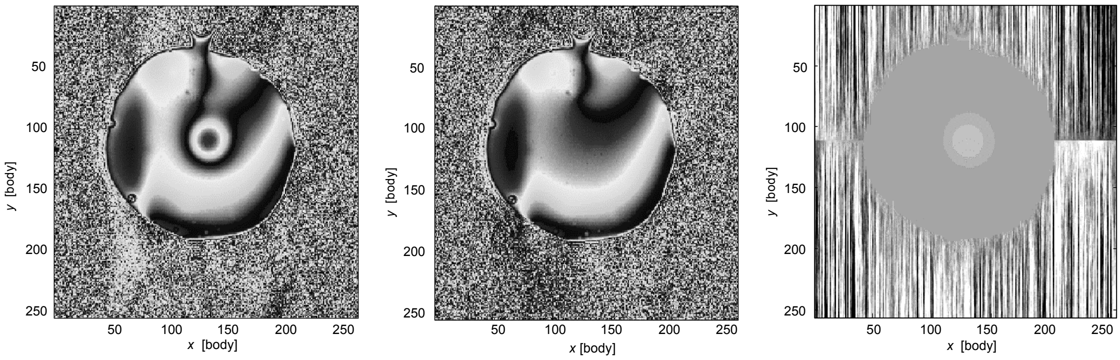


Fig. 3. MR phase images. On the left with the measured coil excited, in the centre the background, and on the left the image weighted by the magnetic induction in the selected layer.

In laboratory coordinates, the image of 256 x 256 pixels corresponds to the size 60 x 60 mm. The spin echo time was *TE* = 5.66 ms. A phase change of *2π* rad in the image corresponds to a change in the magnetic field of $\Delta B = (1/\gamma 0.00566) = 4.15\ \mu\mathrm{T}$, i.e. 0.88 ppm. On the left is the phase image in the case that a current of 5 mA is flowing through the helical coil. In the centre is the phase image of the background (there is no current flowing through the coil). On the right is the difference of images weighted by the change in the basic magnetic filed $\Delta B(xy)$ corresponding to the map of magnetic field in the selected layer.

In Fig. 4, sections through the phase image at point y = 120 pixels are given. On the left for a state when there is a current flowing through the coil, in the centre for the background, and on the right the waveform of magnetic induction of helical coil in the selected layer is given. The choice of time *TE* affects the magnitude of $\Delta B_{(2\pi)}$. In order to avoid the necessity of developing the phase image (removing the phase jumps of 2π) it is of advantage to choose the TE minimal and always shorter than it corresponds to the maximum inhomogeneity $\Delta B(xy)$.

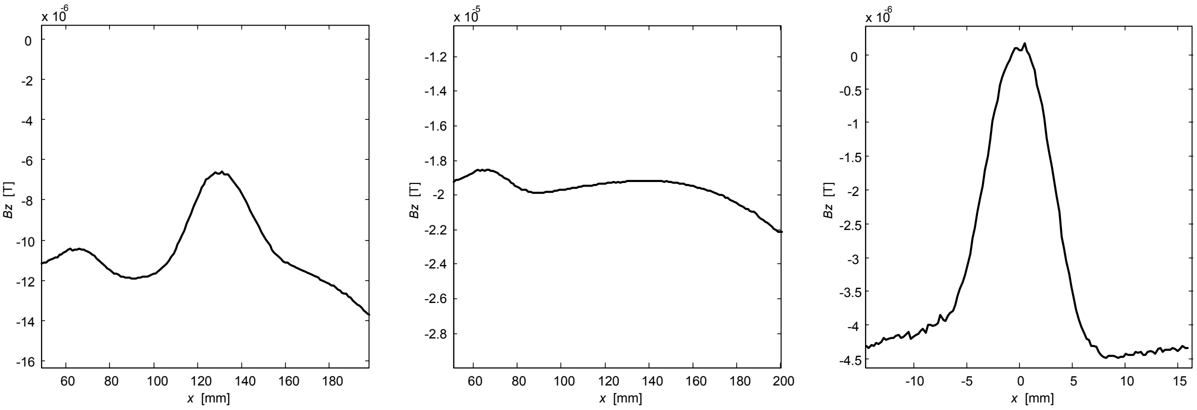


Fig. 4. Sections through phase images at point y = 120 pixels. On the left for a state when there is a current flowing through the coil, in the centre for the background, and on the right the waveform of magnetic induction of helical coil in the selected layer is given.

Fig. 5 gives the same experimental results as in the preceding case but for opposite

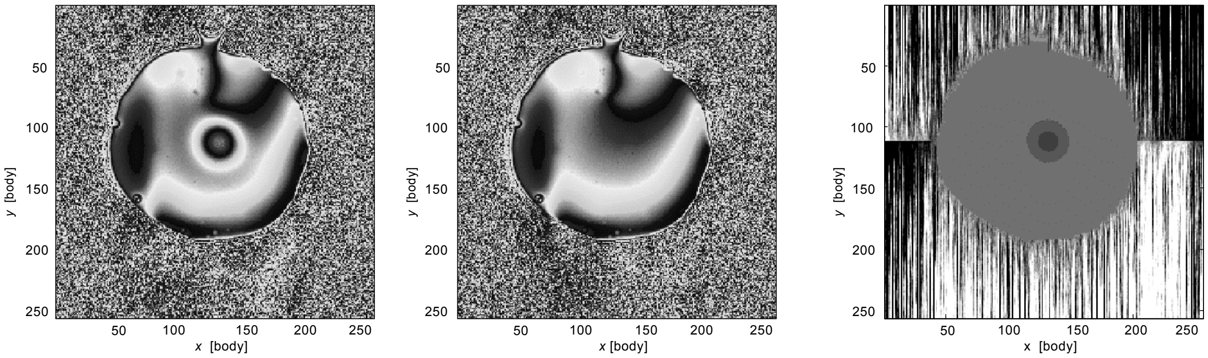


Fig. 5. MR phase images. On the left the measured coils excited with current I = -5 mA, in the centre the background image, and on the right the image weighted by magnetic induction of helical coil.

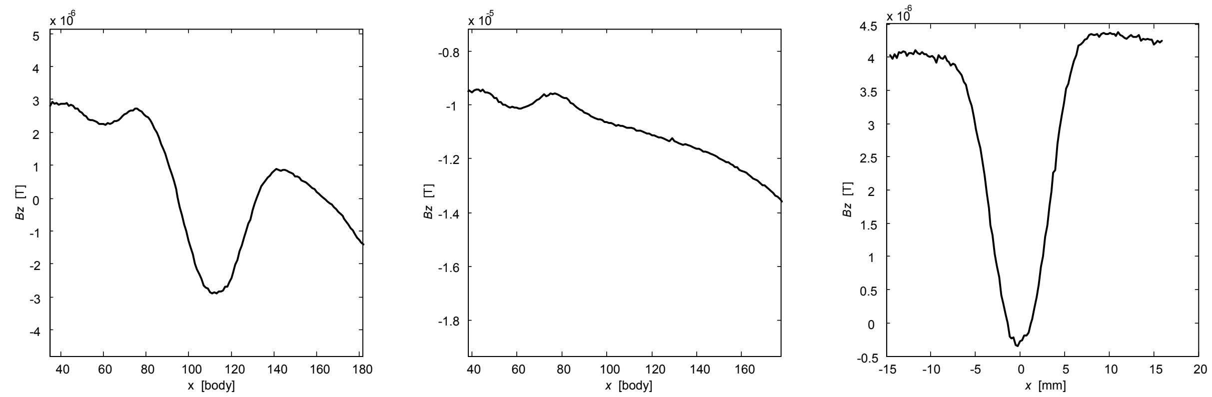


Fig. 6. Sections through phase images at point y = 120 pixels. On the left for a state when current I = -5 mA is flowing through the coil, in the centre for the background, and on the right the waveform of magnetic induction of helical coil in the selected layer is given.

polarity of the current flowing through the helical coil.

It is evident from Figs 3 to 6 that the maximum value of magnetic induction at distance $z = 2.6$ mm from the centre of helical coil is $B_z = 4.4$ T/5 mA while at distance $x \pm 5$ mm B_z decreases to a level of 10 % of the maximum value.

3. Calculation of magnetic induction of helical coil

The waveform of the induction of the magnetic field of helical coil can be calculated simply as the sum of contributions from each individual turn [2]. The calculation can be simplified if instead of the helix a thin circular turn is considered whose diameter corresponds to the mean diameter of one turn. For the magnetic field of a thin circular turn along its axis ($r = 0$, $k = 0$) the radial component is $B_r = 0$, and the axial component of induction is given by the relation

$$B_{z0} = \frac{\mu_0 I}{2}\frac{R^2}{\left(R^2+z^2\right)^{3/2}}. \qquad (5)$$

The total induction of magnetic field will be the sum of contributions from each circular turn

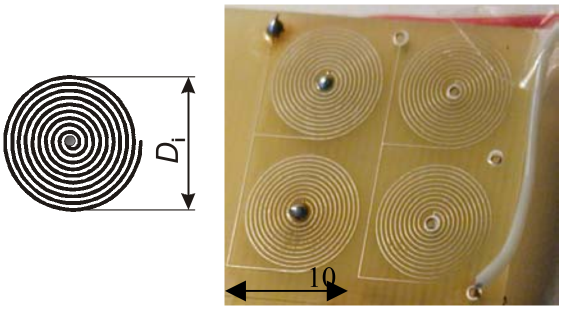
$$B_{z0n} = \sum_{i=1}^{n}\frac{\mu_0 I}{2}\left(\frac{R_i^2}{\left(R_i^2+z^2\right)^{3/2}} + \frac{R_i^2}{\left(R_i^2+(z+a)^2\right)^{3/2}}\right), \qquad (6)$$

where

n	is the number of coil turns,
R_i	is the radius of the i-th turn,
a	is the distance between coil centres (the thickness of printed circuit).

The coil is formed by a pair of identical coils made on a two-layer printed circuit, which are interconnected in the centre, see Fig. 7. The coil has 2x10 turns, and the diameters of individual turns are $D_i = 10$, 9.2, 8.4, 7.6, 6.8, 6.0, 5.2, 4.4, 3.6, and 2.8 mm. The conductor thickness is 0.1 mm. The excitation current was $I = 5$ mA.

For the sake of comparison with the measurement by the MR method we will calculate the magnetic field induction on the axis at distance z from the centre of one of the coils. The axial component of coil induction for $a = 1.5$ mm at distance $z = 2.6$ mm is $B_Z = 4.3$ T/5 mA.



a) b)

Fig. 7. Helical coil a) schematically
b) in four-coil arrangement for testing.

As a check, the magnetic induction was measured using a Tektra gausmeter which operates on the Hall effect principle. Since the minimum range of the magnetometer is 2 mT with 3-digit resolution, it was necessary to supply the helical coil with a larger current, $I = 2$ A, and perform the measurement for a period of 5 seconds. The Hall probe was placed at distance $z = 2.6$ mm. The value measured was $B_z = 1.75$ mT/2 A, i.e. $B_z = 4.375$ mT/5 mA. This measurement carries large errors because the Hall probe cannot be positioned precisely and because of its dimensions (4 x 1.5 mm).

4. Conclusion

The MR method described above was used to measure magnetic induction at a distance of 2.6 mm from the surface of dermatological applicator in a plane perpendicular to its axis. The skin and hypodermis will be in this plane during treatment. The maximum induction value is 4.4 mT/5 mA. For the above design it follows that for current pulses $I = 2$ A a maximum magnetic induction of 1.76 mT can be obtained. A comparison of the results of measuring by the MR method and by the classical magnetometer and the simplified calculation gives a good agreement. The advantages of the measurement by the MR method are the high sensitivity, high resolution (currently up to 0.1 mm), the speed of field mapping, and the possibility of measuring small objects.

Acknowledgment

The paper was prepared within the framework of N°IAA2065201 *project of the Grant Agency of the Academy of Sciences of the Czech Republic.*

References

[1] Blimlich B.: NMR Imagin of Materials. Clarendon Press, Oxford, 2000.

[2] Kulda J.: Magnetické pole silnoproudé elektrotechnice, Academia, Praha, 1974.