# A Novel Method of Measuring the B1 Field Components of an Unsegmented Linear Coil

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**Introduction:** One of the most important characteristics of the transmit or receive coil used in MRI is the distribution of the radio frequency field $B_1$. Inhomogeneity of $B_1$ field leads to changes in signal intensities and causes serious problems with sequences which require accurate application of a particular flip angle (e.g. inversion recovery). MRI-based techniques were used so far for the measurement of $B_1$ maps. The $B_1$ amplitude is computed from the pixel intensities of gradient-, spin- or stimulated-echo images [1-4]. Our approach is different. We propose measurement of the static magnetic field distribution of the rf coil generated by DC current. The method is based on the fact that the $B_1$ field can be replaced by the static magnetic field distribution in lower frequencies where the wave behaviour of electromagnetic field is insignificant and the parasitic capacitances of the considered coil are negligible.

**Materials and Methods:** The spin excitation is performed by the circularly polarized component $B^+_{1xy}(\mathbf{r}) = \sqrt{(B^+_{1x}(\mathbf{r}))^2 + (B^+_{1y}(\mathbf{r}))^2}$ , which rotates in the same direction as spins precess. Method was tested by using a four-turn saddle coil 100 mm in length. Straight conductors of the coil were placed along the tube perimeter evenly. The coil was made of enamelled copper wire ( $\phi = 1$ mm) on a plastic tube with diameter of 75 mm. The self-resonance frequency of the coil was 30.8 MHz. The measured coil was immersed into a plastic basin containing tap water. The main axis of the coil was perpendicular to the static magnetic field $\mathbf{B_0}$. The coil was placed at the centre of the magnet so that measured component $B^+_{1x}(\mathbf{r})$ or $B^+_{1y}(\mathbf{r})$ was aligned with the static magnetic field $\mathbf{B_0}$. The magnetic field distributions were measured without and with DC current (48 mA). Magnetic field deviation caused by the DC current was obtained by subtraction of the magnetic field measured without current from the field measured with the electric current.
The static magnetic field in the axial plane has been measured by a high-resolution spectroscopic imaging technique [5, 6]. The measurements were performed using a 1.5 T scanner (Gyroscan, Philips). A whole body coil was used for excitation and a standard quadrature head coil served as receiver. A spectral bandwidth of 10 ppm and spectral resolution of 0.31 ppm were achieved with 32 image records with a time echo increment, $\Delta$TE of 1.56 ms ($TE_1$ = 10 ms). The slice thickness was 2 mm, bandwidth per pixel 217.1 Hz and flip angle 20$^o$. Image matrix (256, 256), FOV = 160 mm and 192 phase encoding steps led to resolution in the plane of 0.63x0.83 mm. The net measurement time was 10 minutes 15 seconds (1 acquisition, TR = 100 ms). The measured data matrix (512, 192, 32) was processed by 3D FFT after zero-filling to a (512, 256, 512) data matrix. The magnetic field distribution was obtained from the shift of the voxel's water line. The theoretical $B_1$ field distributions were computed using the Biot-Savart law with a software package Mathematica.

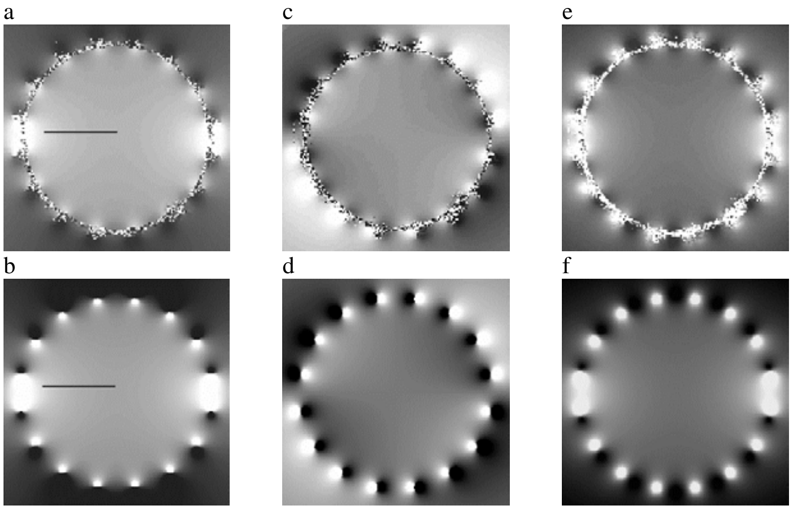

**Fig.1.** $B_1$ field variations of the saddle coil in central axial plane. (a) $B^+_{1x}(\mathbf{r})$ measured, (b) $B^+_{1x}(\mathbf{r})$ theoretical, (c) $B^+_{1y}(\mathbf{r})$ measured, (d) $B^+_{1y}(\mathbf{r})$ theoretical, (e) $B^+_{1xy}(\mathbf{r})$ measured and (f) $B^+_{1y}(\mathbf{r})$ theoretical distribution. Black lines show pixels used for quantitative comparison of the theoretical and experimental values (Fig. 2).

**Results:** $B^+_{1x}(\mathbf{r})$, $B^+_{1y}(\mathbf{r})$, and $B^+_{1xy}(\mathbf{r})$ field variations of the saddle coil are shown in Fig. 1 (middle plane of the coil). Measured magnetic field variations caused by DC current, i.e., after subtraction of the $B_0$ inhomogeneity background are shown in Fig. 1 a, c, e. Figures 1b, d, f show theoretical distributions. The quantitative correlation between the experimental and theoretical values is shown in Fig. 2.

**Discussion and Conclusion:** Figures 1 and 2 show a good qualitative and quantitative similarity between experimental and theoretical $B_1$ distributions. The deviations can be explained by imperfections in the construction of the coil and by size of the voxels. Note that MRI-based techniques [1-4] are able to measure only the "active $B_1$ field", i.e., the component $B^+_{1xy}(\mathbf{r})$ which induces flip angle. Our method is based on the assumption that all electric current is flowing through the coil at operating frequency $f_0$. Applicability of this approach is, therefore, restricted to the lower frequency range and for the coils with small parasitic capacitances. The method will be valid for the unsegmented linear coil without electrically parallel current paths or significant phase shifts within the coil in which (a) the total current path length is less than $\lambda/4$, (b) the wavelength within the sample is less than $\tilde{\lambda}/10$ and (c) the operating frequency of the coil is below self resonance. These conditions are fulfilled by many rf coils operating up to ~20 MHz ($B_0$ = 0.5 T).

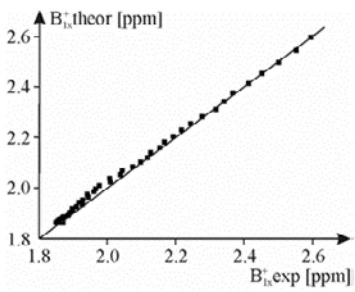


**Fig. 2.** A comparison of the theoretical and experimental magnetic inductions. The values were taken from the pixels depicted by black lines in Fig. 1. Full line represents identity f(x) = x.

**References:** 1. Akoka S at al, Magn Reson Imag 11:437 (1993). 2. Insko EK et al, J Magn Reson Series A 103:82 (1993). 3. Barker GJ et al, Br J Radiol 71:59 (1998). 4. Sled JS et al, IEEE Trans Med Imag 17:653 (1998). 5. Weis J et al, MAGMA 5:201 (1997). 6. Weis J et al, MAGMA 18(6) in press (2005).

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