# B1 and B0 mapping of a micro helix Coil at 9.4T

**M. Mohammadzadeh[1], H. Weber[1], N. Baxan[1], V. Badilita[2], J. Maclaren[1], J. Hennig[1], and D. v. Elverfeldt[1]**
[1]Dept. of Diagnostic Radiology, Medical Physics, University Hospital Freiburg, Freiburg, Germany, [2]Dept. of Microsystems Engineering-IMTEK, University of Freiburg, Freiburg, Germany

## Introduction

The magnetic field produced per unit current, B1, and B0 mapping are important criteria when evaluating the performance of micro coils in NMR microscopy [1]. To increase the coil B1 across the sample and achieve high MR image SNR, RF coils should be placed as close as possible to the NMR sample. Bringing coils close to materials with different susceptibility, results in inhomogeneities in the main magnetic field B0. In turn, this broadens the spectral width and induces artifacts in the image [2]. In this study B1 and B0 maps of a micro helix coil have been measured on a 9.4 T system. The micro coil consists of five turns of gold wire wound on an SU8 cylinder (Fig. 1). It is fabricated in a fully MEMS technology and was used in transceiver mode.

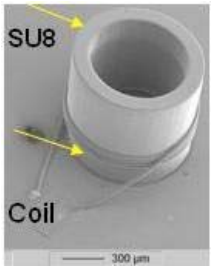


Fig.1: Micro helix coil

## Methods

### B1 mapping

We used the multi flip angle MFA approach to map B1 around a micro helix coil on a 9.4 T system [3]. According to the MFA method, the power of RF transmitter coil was changed by a fixed increment to provide a full period of oscillation to flip the spin angle and thus the MR signal intensity in the area of highest B1 field. Then, for each flip angle, a 3D GE image of a doped water phantom was run for high resolution imaging (matrix: 256×32×16, FOV: 10×1.2×0.4 mm, TR: 1250 ms, TE: 5.08 ms, number of repetitions: 10, number of averages: 4, resolution: 40µm isotropic, imaging time: 7 hours 7 minutes). A signal time course is then produced for each voxel. A B1 map was then calculated by fitting appropriate functions to each voxel signal course.

### B0 mapping

Two 3D GE sequences (matrix: 256×128×32, FOV .6.4×3.2×0.8mm, TR/TE1, TE2 300/7.5,22.5ms, resolution 25µm isotropic) were used to acquire phase Fourier transformed (FT) images at two different echo times, TE1 and TE2. The frequency difference, $\Delta\omega_0$ (in Hz), is proportional to ΔB0 (in mT). Therefore B0 maps are produced by [4]:

$$\Delta\omega_0 = \gamma\Delta B_0 = \frac{\Delta\Phi}{\Delta TE}, \qquad (1)$$

where ΔΦ is the phase difference for each voxel, calculated from the two MR images, ΔTE is the difference between two TEs, which was set to 15 ms in this study. Unwrapping and image subtraction of two phase images were performed in Matlab.

### Imaging system

This project was implemented on a 9.4 T Biospin 94/20system(Bruker Biospin, Ettlingen, Germany). Doped water solution was filled in SU8 and sealed to prevent evaporation during the long scan time. During the test, the micro coil's axis was perpendicular to the $B_0$ field and to get high resolution images only 3D sequences were used.

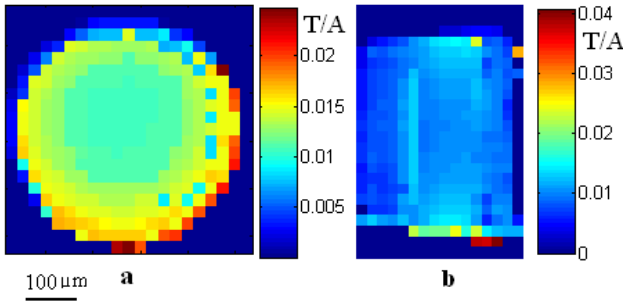


Fig.2: 2D B1 map of the Micro helix coil in a: Coronal slice, b:Sagittal slice.

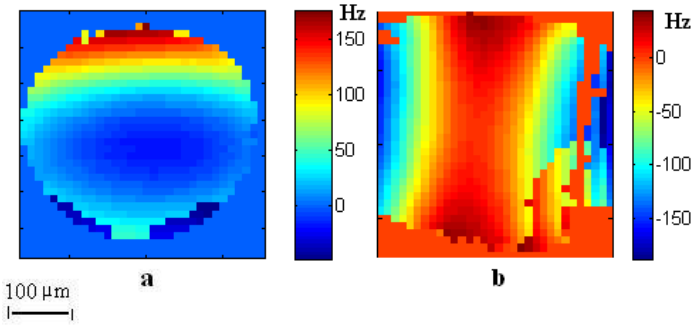


Fig.3: 2D B0 map of the Micro helix coil in a: Coronal slice, b: Sagittal slaice.

## Results and Dicussions:

The MFA method successfully provides B1 maps of the micro coil in coronal and sagittal slices. Figure 2a,b show a uniform area inside doped water phantom around SU8 axe. In addition they indicate that signal intensity near the coil's wire increases. The reason for this is that the normal vectors of the coronal slice and loops of the helix are not parallel. Color bars indicate a maximum of 0.04 T/m variation for the B1 across the sample, while this parameter changes in the centre (green area) less than 0.005 T/m.

Figure 3a,b represents 2D B0 maps of the coil in coronal and sagittal slices, respectively. Color bars indicate that the maximum Δω0 changes across the sample is 150 Hz. However in the centre it reached less than 10 Hz.

## Conclusion:

Results show that the micro helix coil produces high SNR and acceptable in homogeneity, such that we produced high resolution images of a nanoliter sample with a resolution of 25 µm. In addition this indicates that by improving the susceptibility matching of the coil structure and using array coil design it is possible to image regions of size 1 mm across at a resolution of micrometers, which nowadays is not possible using standard MRI systems.

References & Acknowledgments

[1] Hoult., NMR 1978, 12: 41-47; [2] Subramamian et al., JMR 1998, 133: 227-231; [3] Weber et al., ISMRM 2009, #370; [4] Chen et al., Med. Phys.2006, 33(11): 4299-4306;.

The authors thankfully acknowledge the European Union for financial support (FP6-NEST-2004: Micro MR)