Performance of Inductively Coupled RF Coils as Reference Markers under Various Conditions at 1.5 T

R. Trampel[1], H. Busse[1], W. Gründer[2], M. Moche[1], and T. Kahn[1]

[1]Diagnostic and Interventional Radiology, University Hospital Leipzig, Leipzig, Germany, [2]Institute for Medical Physics and Biophysics, University Leipzig, Leipzig, Germany

Introduction

Various layouts for guiding minimally invasive interventions with the help of magnetic resonance (MR) imaging have been successfully deployed in the past decade. Using a conventional closed-bore MR scanner requires the co-registration of the patient and instrument position outside the magnet with the scanner's coordinate system. Micro coils connected with the hardware of the MR scanner may be used as MR markers (1) but suffer from safety problems due to the electrical leads (2). Inductively coupled radio-frequency (RF) coils are not object to such problems and still have a superior contrast-to-noise ratio (CNR) than other passive markers (3-5). The CNR depends on coil-related parameters (resonance frequency and quality factor Q of the marker coils) and on experimental parameters such as the used flip angle and the distance of the marker coils from the isocenter of the magnet. Therefore, our goal was to investigate the performance of self-developed, inductively coupled RF coils as reference markers under a broad range of experimental conditions at 1.5 T.

Method

The inductively coupled RF coils consisted of four turns of copper wire (Ø 0.3 mm), had an approximate size of 2 x 3.5 mm (length, outer diameter), and were tuned to a resonance frequency of 63.9 MHz using a 100 pF capacitor. The Q factors of the marker coils were between 80 and 90. The coils were wound around a plastic tube filled with a T_1-shortening contrast agent. To allow a well defined positioning of the coils they were mounted on 2x2 LEGO bricks and attached to a LEGO plate. All MR experiments were performed on a 1.5 T whole-body MR scanner (Magnetom Symphony, Siemens Medical Solutions, Erlangen, Germany). As a reference a water-filled plastic bottle (Ø 11.5 cm) was utilized. A gradient echo fast low angle shot (FLASH) sequence (TE=6.9 ms, TR=20 ms, FOV=30 x 30 cm^2, matrix: 512 x 512, slice thickness ST=30 cm, bandwidth BW=220 Hz/Px) and a true fast imaging with steady state precession (TrueFISP) sequence (TE=2.85 ms, TR=3651 ms, FOV=30 x 30 cm^2, matrix: 512 x 512, ST=20 cm, BW=220 Hz/Px) were used. For both sequences coronal, sagittal and axial images were acquired. The performance of the coils in dependence on the flip angle and on the position within the scanner was investigated. Quantitative evaluation involved the analysis of the mean signal intensity of three different kind of ROIs: (1) *coil* ROIs at the positions of the coils, (2) a larger *background* ROI of the phantom, and (3) a *noise* ROI in the periphery.

Results

Figure 1 shows the variation of the signal intensities of the ROIs as a function of the flip angle in a coronal image acquired with a FLASH sequence. For FAs in the range of 0.5-1.1°, the contrast between coils and background (bottle) goes through a maximum. For the TrueFISP sequence the dependence is similar (maximum CNR for FA ~ 1-2°) but the absolute signal intensity of the marker coils is more than two times higher than for the FLASH sequence. However, the typical signal bands of balanced steady-state free precession sequences may interfere with the detection of the coils especially at small flip angles (6). The results are similar for axial and sagittal images. In Fig. 2 the signal variation in a sagittal image as a function of the position along the *z*-axis of the magnet for the TrueFISP sequence is shown. Again, the results for the FLASH sequence and for coronal/axial slice orientations are similar. The optimum position of the coils is in a region of about ± 10 cm around the magnet's isocenter. However, despite the high Q factor of the coils, the signal is still well above the noise and background level in a region of about ± 20 cm around the isocenter (Fig. 2). Similar results were obtained if the coils were moved along the *x*- and *y*-axes which makes the coils applicable in a volume of 40 x 40 x 40 cm³.

Discussion and Conclusion

The performance of inductively coupled RF coils as MR markers under a broad range of experimental conditions was investigated. The great advantage of such a design is the absence of connecting wires which are a source of potential hazards. Furthermore, conventional imaging sequences can be used for the localization of the markers. Whereas the CNR is sufficient in a region of ± 20 cm around the isocenter (Fig. 2) non-linearities of the magnetic field gradients might cause localization errors at the extreme positions. Appropriate correction algorithms may be used to overcome such problems. However, phantom experiments (not shown here) demonstrated that in the region of about ± 10 cm around the isocenter (where an excellent CNR was achieved) the maximum 1D localization error was below 1 mm. Hence, in combination with an image based position detection algorithm, the self developed RF marker coils could be used for an automatic patient registration.

References

1. Dumoulin CL et al., Magn Reson Med 1993;29:411-415. 2. Wildermuth S et al., Cardiovasc Intervent Radiol 1998; 21:404-410. 3. Burl M et al., Magn Reson Med 1996;36:491-493. 4. Flask C et al., J Magn Reson Imaging 2001;14:617-627. 5. Quick HH et al., Magn Reson Med 2005;53:446-455. 6. Mekle et al., J Magn Reson Imaging 2006;23:145-155.

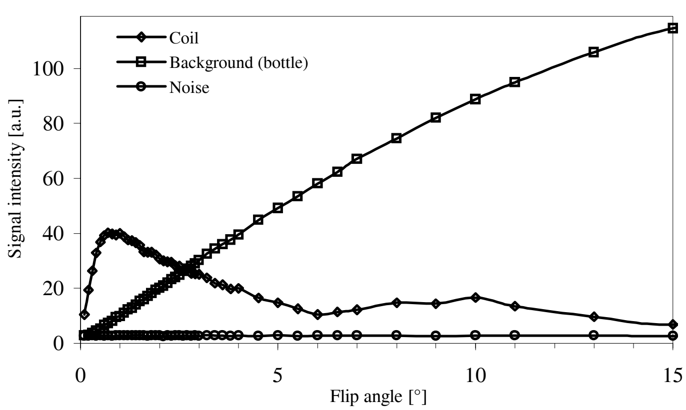

Fig. 1: Variation of mean signal intensities in coil, background and noise ROIs in a coronal image obtained with a FLASH sequence as a function of flip angle.

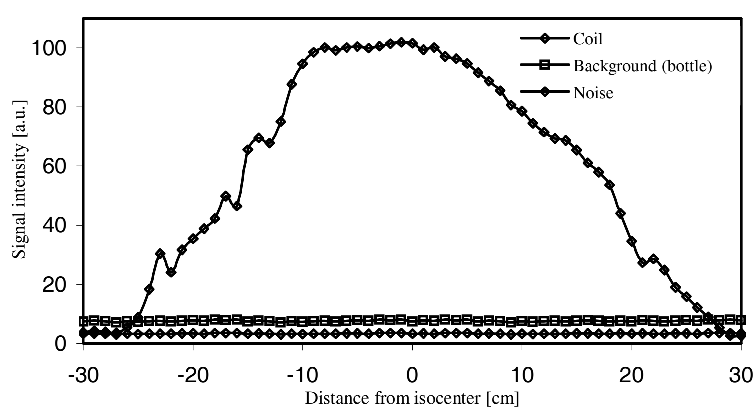

Fig. 2: Variation of signal intensities in a sagittal image obtained with a TrueFISP sequence (FA=1°) as a function of position along the *z*-axis of the magnet.