Current Sheet Antenna Array - a transmit/receive surface coil array for MRI at high fields

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Introduction: Phased arrays of loop surface coils or planar strip arrays [1] have been introduced to acquire conventional and parallel spatial encoded MRI imaging in the past. So-called current sheet antennas (CSA) have been commonly used as hyperthermia applicators applying the RF electric field to heat body tissues. The main problem impeding a wide use of these applicators is the low penetration depth of the electrical field. This yields in low efficiency of the applicators for heating tissues. A basic analysis guided by computer simulations shows the absence of the normal component of the electrical field. Additional modifications for shielding electrical fields result in an excellently performing surface coil which is applicable as single coil or in combination with multiple CSA-coils for array imaging, e.g. parallel imaging.

Methods: CSA coils have a boxed conducting shape, open sides and a slot at the opposite side from the VOI of the coil (see figure 1). This slot is bridged by capacitors with fixed capacity and a tuning capacitor to adjust the frequency at 125.3 MHz. A gamma-feeding is implemented in the inner volume of the coil for excitation. The CSA coil was fabricated with normal FR4 base material coated with 8µm copper with following dimensions: width w=8 cm, length l=16 cm and height h=3 cm. For the prototype coils four high-Q capacitors (2x 5.6pF, 2x 3.9pF) and a variable tuning capacitor (Temex 1-10pF) are used. Initial tests with a single coil have been done on the bench and in a 3T MR system. The single elements are arranged as a four element transmit/receive array on Perspex mounting usable for head imaging with an inner diameter of 25 cm. The array elements are not separately decoupled by decoupling or feeding networks! All MR-measurements were performed on a 3 Tesla whole body MR-scanner (MEDSPEC30/100 Bruker BioSpin MRI GmbH, Ettlingen, Germany). The system has four fully equipped transmit/receive channels which can be independently driven during the experiments. The individual array elements are connected via transmit/receive switches, each equipped with a LNA- preamp (gain 27dB and NF < 0.8dB). The transmit path of the coil is configured with an additional circulator for decoupling the coil itself completely from the transmitter. This is important when the array elements are individually driven beyond the naturally decoupled modes, e.g. birdcage mode. This means for a four element array to drive the array elements with the same amplitude and phase shifts of 0°, 90°, 180° and 270° respectively. The array coil was first loaded with a cylindrical Perspex phantom (i.d. = 20 cm, l = 20 cm) filled with agarose gel containing 1.33 g/l NaCl and 0.66 g/l CuSO4 to achieve an electrical conductivity of 0.33 S/m and a permittivity of 76 (at 100 MHz and 28°C) similar to the in-vivo case. As the intensity distribution of a gradient echo image with low flip angles of a homogeneous phantom is proportional to $|B_1^{(+)}B_1^{(-)}|$, we acquired the images with a standard gradient echo with flip angles α <10°. Comparisons of measurements and simulations of the B1-field and SAR were done with methods discussed in [2, 3].

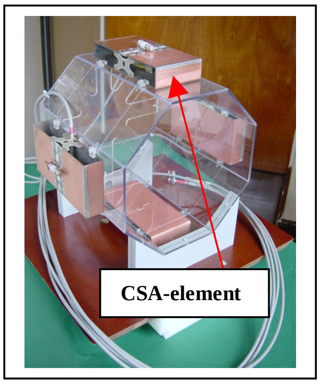


Figure 1

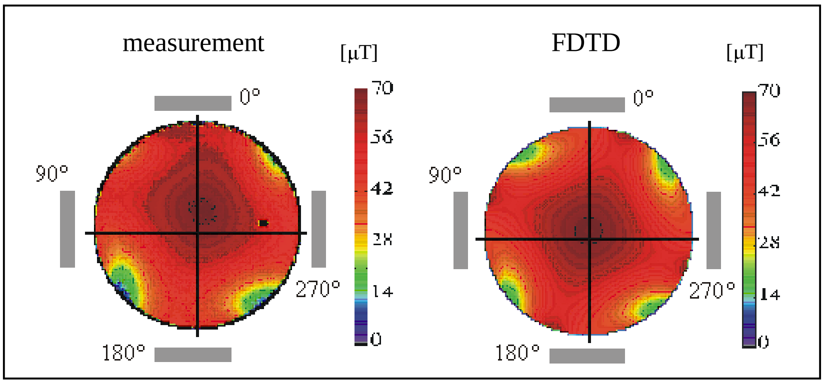
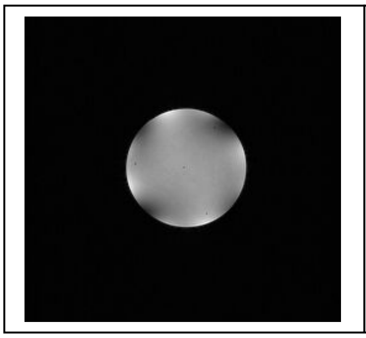
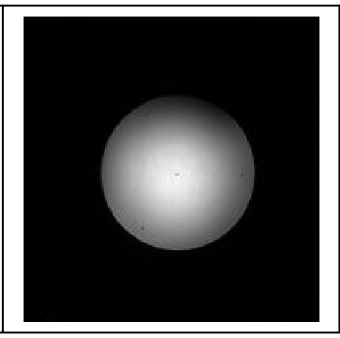


Figure 2





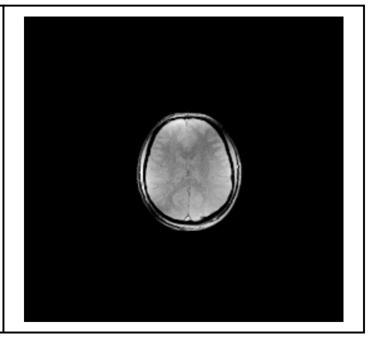


Figure 3 **Figure 4** **Figure 5**

Results: Figure 1 shows a photograph of the 4-element array coil used in this study. On the bench a single array coil element shows an excellent performance in the loaded and unloaded case (Q_{ul}=180 Q_l=80). A comparable single loop surface coil shows for the same phantom less performance (Q_{ul}=180; Q_l=40). The comparisons of the B1-field measured in the MR-system show a good agreement with FDTD data both for the $|B_1^{(+)}|$ distribution (figure 2) and the absolute center value of $|B_1^{(+)}|$ = 71µT (at 1kW per coil). Additional experiments with different amplitudes and phases for manipulating the B1-field distributions show exactly the results predicted by simulations. A comparison between a standard birdcage head coil and the CSA array coil with a cylindrical Perplex phantom was done; figures 3 and 4 show transverse images of the phantom. Figure 3 displays a sum-of-squares combination of the images of the individual coil elements. One can clearly see a "dielectric resonance" like enhancement of image intensity for the birdcage whereas the intensity distribution for the CSA array is much more homogenous (figure 3). Initial in-vivo experiments show the same B1-field characteristics as in the phantom experiments (figure 5).

Discussion and conclusions: The high potential of using CSA-array coils for high field MRI is demonstrated. This opens new approaches for transmit/receive array coils at high and ultra-high fields. Next steps will be driving the coil elements with different amplitudes and phases thus allowing a straight forward adjustment of the B1-field homogeneity. Increasing the number of array elements will yield additional benefit regarding SNR and homogeneity.

References: [1] A. Kumar, P. A. Bottomley; Proc. Intl. Soc. Mag. Reson. Med. 321 (2002); [2] F. Seifert, S. Junge, H. Rinneberg; Proc. Intl. Soc. Mag. Reson. Med. 321 (2002); [3] S. Junge, F. Seifert, H. Rinneberg; Proc. ESMRMB 424 (2002)