A New Helmet Coil Concept Using Strip Lines

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Introduction

Compared to the standard birdcage head coil, an anatomically shaped helmet coil provides a better filling factor and an improved SNR in the superior parts of the brain [1]. In previous designs, the coil was optimized for high sensitivity in regions such as the primary motor cortex. However, the trade-in was a large B_1-gradient along the z-axis. In order to permit imaging or spectroscopy studies in deeper regions of the brain with improved sensitivity, we used strip-line (SL) technology to create a novel helmet coil concept without a circular base to reduce the energy deposition within the tissue and a reduced B_1-gradient along the z-axis. In addition, it is stabilized with respect to different loading conditions, easier to handle and (due to its open design) more convenient for paradigm presentation in fMRI studies. It should be noted that this coil concept is also useful for frequencies exceeding 125 MHz (e.g., for human brain imaging at 7 T).

Coil design

The SL technology used for the new coil design (Fig. 1) consists of thin strip conductors and a ground plane both separated by a low-loss dielectric material [2]. The most important advantage of a resonator based on SL technology is the quasi-transverse orientation of the electromagnetic field.

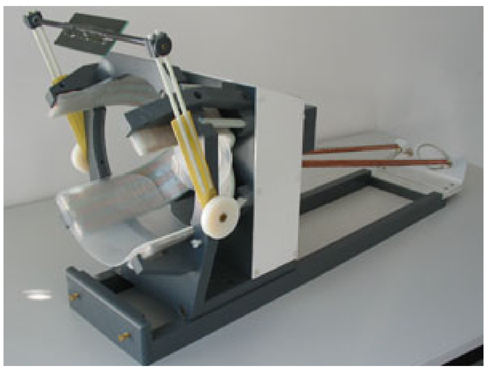

Figure 1: Design of the new helmet coil.

The dielectric material (in our case polypropylene of thickness 5mm) has a dome-like shape with 23-cm inner diameter and 18-cm height. The overall design consists of four segments rotated by 90° around the z-axis. Three 0.9 cm wide SL's with a characteristic impedance Z_0 separated by a gap of 1.6 cm are attached to the inner side of each segment. To reduce eddy currents, both the SL's and the ground planes on the back side of each segment are made from sufficiently thin self-sticking copper foil (thickness: 18 µm). At the lower end (*i.e.*, near the neck) the SL's are terminated by a short (load impedance $Z_L = 0$). In this case, the the load reflection coefficient is $\Gamma_L = -1$ and standing waves are created along the line with a current maximum at the termination point. According to transmission-line theory, the shorted line behaves like a parallel resonant circuit at frequencies in the vicinity of quarter wavelength ($\lambda/4$). The phase constant β multiplied with the actual length l of the SL (l = 23 cm) is then $\beta\, l \sim \pi/4$, and the input impedance (Z_{in}) is inductive because $Z_{in} = j\, Z_0 \tan \beta\, l$ is positive. Trimmer capacitors connected parallel to the input of the SL tune the coil to 125.4 MHz. Additional variable capacitors connected in series with the inner conductor of the semi-rigid cable to the generator match the impedance at the tuning capacitor to that of the semi rigid cable (50 Ω). It should be noted, that the current distribution along the line and the field profile might be optimized by using a dielectric material with a different dielectric constant.

Results and Discussion

Experiments were performed on a 3-T Bruker MedSpec 30/100 system. Using a cylindrical water phantom (length 37 cm, diameter 16 cm) aligned along the symmetry axis of the coil, the reference power to obtain a 90° flip-angle was mapped depending on the position along the z-axis (Fig. 2). Initial *in vivo* applications in human volunteers in comparison to results obtained with the original helmet coil verify that the new design achieves sufficient image contrast almost in the entire brain. Examples of corresponding axial T_1-weighted MDEFT images for both the original helmet coil and the new design from the same subject are shown in Fig. 3. The new helmet coil provides similar sensitivity in the upper brain regions. At the position of the eyes (about 12 cm distant from the top) a 70 % gain in the B1 field is obtained, cf. Figs. 2 and 3. In addition, the in-plane B_1 variation at this position is significantly lower (Fig. 3). Whereas only the superior parts of the brain are visible with sufficient SNR and contrast with the original helmet coil (A, B), high-quality images of deeper structures are also obtained with the new coil (C, D). The modified design provides sufficient space for use of additional audiovisual stimulation devices to be used in fMRI and appears, hence, well suited for anatomical and functional studies. Another advantage is a substantially reduced stray field at the bottom of the coil, which can be exploited for continuous arterial spin labeling (CASL) experiments with a passively decoupled second coil for labeling the blood in the carotids [3].

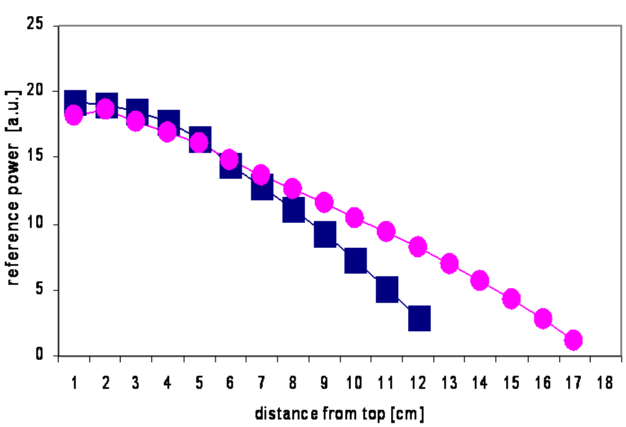

Fig. 2: *Reference power* (squares: original helmet coil; circles: new coil).

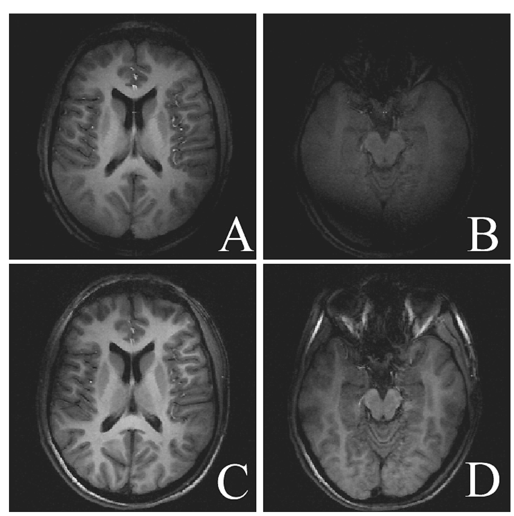

Fig. 3: T_1-weighted MDEFT images (T_R 1.3 s, T_E 10 ms, FOV 192 mm).

References

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