A Four-Channel Transceive Phased-Array Helmet Coil for 3 T

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Indroduction

In many imaging or spectroscopy applications, surface coils are used to improve the signal-to-noise ratio (SNR) in peripheral regions close to the coil [1]. However the trade-in is a strong inhomogeneity of the radiofrequency (RF) field, especially if the same coil is used for transmit and receive. This problem is often tackled by using a volume-coil transmitter decoupled from a surface-coil receiver. Another common strategy to improve the sensitivity and obtain sufficient SNR from deeper-lying regions is to use a phased-array receiver [2] in combination with a volume-coil transmitter, typically the body coil. However, for some MR scanners, such as dedicated head scanners or ultra high-field systems (operating at fields above 3 T) whole-body transmit RF coils are normally not available due to design challenges and specific absorption rate (SAR) issues. It would, hence, be desirable to have a single head coil that combines the advantages of a transmitter of sufficient homogeneity and a multi-channel receiver. Aim of the present work was to design a 4-channel transceiver array coil for MRI and MRS of the human brain at 3 T, which may also be used as a prototype model for higher frequencies.

General coil design

To achieve a high filling factor, a dome-like helmet coil design (24-cm inner diameter, 20-cm length; Fig 1) with strip-type transmission-line resonators was used for the construction of the transceiver array. It consists of two thin metal strips (width 7 cm; thickness 10 µm) separated by a low-loss dielectric substrate (15-mm thick polypropylene) and terminated with a short at the end opposite to the load [3]. The electrical length of the strip-line (SL) resonator was equivalent to a quarter wavelength. Upon sinusoidal excitation, standing waves are created along the SL with a current maximum at the short. Each SL resonator was tuned with a shunt capacitance and matched to 50 Ω with a series capacitance. To reduce losses caused by the electric field a balanced tune/match network in combination with a quarter-wavelength balun was used to fit the unbalanced coaxial cable to the balanced load of the coil. The balun was custom-made using shrink tubes (polyolefine, ε ~ 2,2) to achieve an outer diameter of 8 mm around the semi-rigid transmission line of 3.6-mm diameter. For the outer jacket, we used self-sticking copper foil (70-µm thick). This sleeve and the outer jacket of the semi-rigid transmission line form a shortened quarter-wave line, tuned to the resonance frequency of the coil. The T_X power was split into the four SL resonators with phases and amplitudes to produce a circularly-polarized field, $\mathbf{B_1}$, inside the helmet coil. In particular, a 180° power splitter was used in combination with two 90° splitters (Stark Contrast, Erlangen, Germany) to produce phases of 0°, 90°, 180°, and 270°. Power was controlled by a T/R switch for each channel that used actively switched PIN diodes to provide sufficient isolation between transmitter and receiver. To minimize the mutual coupling among the array elements during reception, preamplifiers with high input-impedance were used for each channel.

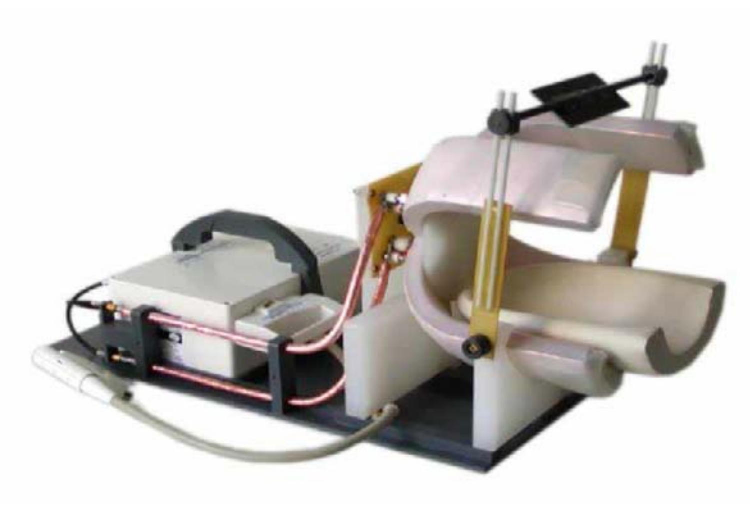

Fig. 1. *Design of the transceive helmet array*

Results and Discussion

To optimize the design, finite-element calculations of the $\mathbf{B_1}$ distribution were carried out with the software package HFSS 9 (Ansoft, Pittsburgh, PA) and analyzed with respect to the efficiency and the occurrence of unwanted hot spots. In axial slices, the numerical results yielded a high degree of circular polarization and a $\mathbf{B_1}$ homogeneity that was comparable to that of a birdcage coil. A small gradient was obtained along the *z*-direction. Experimental results were obtained on two 3-T scanners (Bruker MedSpec 30/100 and Siemens MAGNETOM Trio) with a cylindrical water phantom (length 37 cm, diameter 16 cm; composition 1.24 g $NiSO_4 \cdot 6\ H_2O$ and 2.62 g NaCl per 1000 g water). The $\mathbf{B_1}$ distribution imaged by the double-angle method [4] was consistent with the numerical results. The linear coil mode was used to study the transmit properties from all individual elements of the array (Fig. 2) indicating a high degree of symmetry. Figure 3 shows an image aligned perpendicularly to opposite SL resonators. Again, the experimental results were consistent with the numerical studies. The SNR obtained with the phased-array coil was about 130% of that achieved with a commercial birdcage head coil.

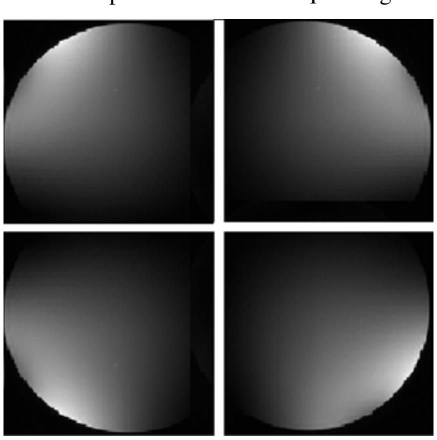

Fig. 2. *Transmit pattern in the linear mode.*

Fig. 3. *Image measured at -45° (sag>cor).*

Conclusion

The helmet array coil prototype can be used more conventionally in receive-only mode or as an array transceiver if a large volume transmit coil is not available. This could find important applications with parallel imaging, especially if more than four channels are integrated.

References

[1] Edelstein WA et al. SMRM, 4th Annual Meeting, San Francisco, 1985, p. 964-965; [2] Roemer PB et al. Magn Reson Med 1990, 16: 192-225; [3] Driesel W et al. Proc. ISMRM 2005, 13: 948 (2005); [4] Stollberger R, Wach P, Magn Reson. Med 1996, 35: 246-251.