A transceive RF coil for imaging tissue specimen at 3T based on PCB design

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Target Audience: Researchers interested in imaging small-animal organs or tissue specimen (3-6cm), RF engineers, MRI physicists

Purpose. Goal of the project was to design and build a transceive (TxRx) radiofrequency (RF) coil for imaging *post-mortem* or *ex-vivo* tissue specimen or small-animal organs on a human-scale MR scanner. Besides high signal-to-noise ratio (SNR), high homogeneity and stability of the magnetic RF field, B_1^+, a major objective was to minimize RF heating of the sample during prolonged imaging experiments.

Methods. Two perpendicular Helmholtz coil pairs were combined to obtain a quadrature TxRx coil to exploit the lower power requirement (3 dB) as compared to a linear coil. To achieve a high degree of accuracy and standardization during coil fabrication, the whole three-dimensional (3D) design was made from printed circuit board (PCB) material and required three different types of boards. The loop shape of the Helmholtz coils was quadratic. Venting slots and windows were integrated in the 3D design for air circulation. To provide access for sample positioning inside the coil, the front part was removable (Fig. 3). Eight non-magnetic brass screws reproducibly provided electrical contact without degrading the homogeneity of the static magnetic field. A rough estimation of loop inductances was obtained for one Helmholtz pair by means of FastHenry [1]. Based on this result, a still simplified 3D model (Fig. 1) was created for simulations with HFSS v11 *(Ansys, Pittsburgh, PA).* It contained the four copper loops, each with two ports, FR4 PCB basic material and different phantom loads. The resulting *S*-parameters were imported into LTSpiceIV *(Linear Technology, Milpitas, CA),* enabling very fast calculation (order of seconds) of the influence from different circuitries and part values.

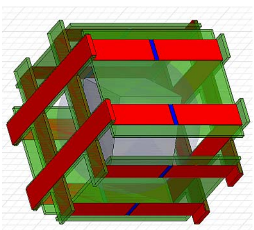

Fig. 1. HFSS model (red: Cu loops; green: FR4; blue: ports; gray: load).

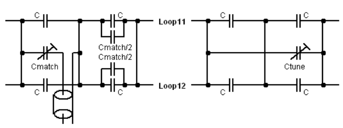

Fig. 2. Matching (left) and tuning (right) circuits of one Helmholtz pair.

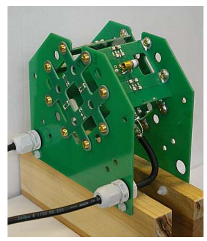

Fig. 3. Final coil. Screws are re-movable to access interior space.

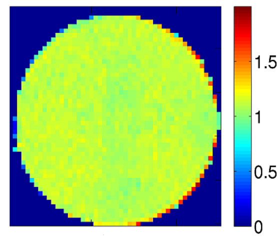

Fig. 4. Normalized B_1^+ map (axial) of the final coil measured using a sample of agarose gel.

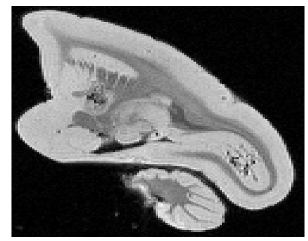

Fig. 5. *Post-mortem* marmo-set brain (3D FLASH; TE/TR 30/300 ms, α 60°, 200µm isotropic nominal voxel size).

The final circuit is shown in Fig. 2. Parallel connections between the loops (66×66mm²; 10mm wide copper traces, 35µm thick) of one pair were made by short copper traces. Each loop was pre-tuned to 125 MHz by fixed capacitors *(C=33pF; 2%, 1111 SMD footprint, 152 CHB series, Temex Ceramics, Pessac, France)*, connected in series to ensure balanced feeding. An additional trimmer capacitor *(55H01, Johanson, Boonton, NJ)* and the feeding coaxial cable were placed exactly midway between the loops. With a matching capacitance, C_{match}, of a few pF, the feed port is already roughly balanced, which was improved by two additional capacitors, each with half the average value of C_{match}. A board with equivalent layout opposite to the feed port carried the tuning circuit.

For imaging experiments at 3T *(Medspec 30/100; Bruker BioSpin, Ettlingen, Germany),* tissue samples or agarose gel were embedded in acrylic spheres of 50 mm diameter, filled with Fomblin.

Results and Discussion. Due to the perpendicular arrangement, the inductive coupling between the Helmholtz pairs can be neglected as long as the currents in both loops of one pair are equal. The capacitive coupling, caused by the unavoidable mutual capacitances of the two loop pairs, is more difficult to treat. To some extent, holes (8mm diameter) in the PCB at the crossings of the copper traces are reducing these capacitances without penalty, because the current density is higher at edges of copper traces [2]. In case of equal effective capacitance at the opposite ends of the loop pairs, the voltages across the mutual capacitances are small and compensating each other. As a consequence the capacitive coupling is load dependent; that is, a higher load requires higher Cmatch and accordingly smaller Ctune. With only one Helmholtz pair and a well-balanced feed port no significant coupling to the coax cable was detectable and, therefore, no extra sheath wave trap was needed.

In CP mode, achieved by connecting both Helmholtz pairs to a 90° hybrid with two coax cables of equal length, tuning and the coupling between both pairs depended on cable length. In addition, some degree of "hand effect" was obtained. These effects were most likely caused by a high-Q Lecher-line type resonator formed by the two coax cables, excited in differential mode. To verify this hypothesis, we increased the damping of this parasitic resonator. Connecting both cable shields by two lumped resistors (similar to Ref. [3], where distributed resistance was described) of 47 Ω (not related to 50Ω cable impedance) at approximately 150 mm and 400 mm from the coil achieved sufficient damping. Again, no extra common-mode trap was required.

Bench-top experiments yielded an unloaded Q of 350 and an isolation between both Helmholtz pairs by more than 20dB. Even during scanning sessions of several days, no detuning was observed. Upon RF heating for 1 hour (10ms Gaussian pulses; $\gamma B_{1,max}$ = 18850 rad/s, TR 30ms), the core temperature inside an agarose gel phantom increased by 6K, compared to 16 K obtained with a single, linearly polarized Helmholtz coil pair. Fig. 4 shows the normalized B_1^+ distribution obtained with a double-angle technique demonstrating good homogeneity. A high-resolution imaging example obtained from a post-mortem marmoset brain is given in Fig. 5.

Conclusion. Using a PCB-based design, inexpensive, reliable, and reproducible 3D coil structures are obtained by appropriate arrangement of the 2D boards. An advantage is a submillimeter precision for the geometrical structure, which is hardly achievable for handcrafted coils. The same principle also simplifies decoupling of multi-channel coils. Dielectric losses of FR4 are manageable with convenient capacitor values, as demonstrated by the obtained Q value. Connecting resonant circuitry by simple non-magnetic brass screws did not lead to additional problems. Differential-mode sheath waves can be easily and aperiodically suppressed by adding a few resistors.

References. [1] http://www.rle.mit.edu/cpg/research_codes.htm [2] J. Mispelter, M. Lupu, A. Briguet. NMR Probeheads for Biophysical and Biomedical Experiments. Imperial College Press, London, UK, 2006; p. 280. [3] E.B. Boskamp et al. Broadband damping of cable modes, Proc. ISMRM 20: 2691, 2012.