SAR consequences of optimization strategy for a 7T RF transmit loop array in CP mode

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Purpose: There are no systematic SAR simulation data for a loop-based array in circular polarization (CP) mode, loaded by a realistic head model, as a function of known array transmit optimization strategies. Furthermore, it is difficult to compare literature SAR values versus transmit performance, because data for $\boldsymbol{B_1}$+ field versus entire head SAR or 10 gram average SAR ($\boldsymbol{SAR_{10g}}$) are rarely reported. Most recent reports have focused on SAR optimization for parallel transmission and RF shimming. While these are hot topics, for many practical applications so far the workhorse for driving a multi-channel transmit RF coil has been a single power amplifier followed by a power splitter and phase shifter. In our analysis two classes of transmit optimization strategies are included: optimization of array geometry (array size, distance of array to load, etc.) and tuning/matching/decoupling.

Method: In the simulation, for array type #1, eight rectangular planar resonance coil elements were mounted on rectangular acrylic supports, assembled to give an octagonal cross section with extreme dimensions of 230 mm by 255 mm. The single loop elements were simulated with widths 85 and 75 mm, and lengths 85, 100, 120, 150 mm. RF array type #2 comprised 8 channels with identical rectangular loops (length 100, 120 mm, angular size 22.5, 27.5, 32.5, 37.5 degree), mounted on a cylindrical acrylic former with diameter ranging from 220 mm to 300 mm. The commercially available Rapid BioMed 7T array [1] was also simulated. Our investigation used RF circuit and 3-D EM co-simulation [2]. All arrays were excited in CP mode, applying 1W power to each port (array transmit power - $\boldsymbol{P}_{\mathbf{transmit}}$=8W), with a sequential 45 degree phase increment. The SAR and the maximum of 10 gram average SAR ($\boldsymbol{SAR_{10g}}$) was calculated using an in-house Matlab procedure with a 1x1x1 mm^3 mesh [3]. For a given geometry, arrays were optimized a) by capacitor and inductor based decoupling networks placed at different positions, b) minimization of power reflected by the entire coil ($\boldsymbol{P}_{\mathbf{array\_refl}}$) c) by ideal Cartesian feedback simulated by using 1A current courses with a sequential 45 degree phase increment placed across each element input, followed by rescaling of the data to ensure that $\boldsymbol{P}_{\mathbf{transmit}}$=8W. The loads utilized were the Ansoft human body (head, shoulders, torso) with different scaling factors: head #1 with scaling X=0.9, Y=0.9, Z=0.9 simulating a small head, and head #2 with scaling X=0.95, Y=0.975, Z=0.9 simulating a large head.

The scanner gradient shield (with diameter of 683 mm and length 1200 mm) was always included in the numerical domain for simulation of unshielded and shielded arrays. The entire data base includes 64 different array geometries, each optimized in post-processing in at least 3 different ways. We analyzed the power balance as well as the transmit magnetic field performance and inhomogeneity (calculated as the ratio of standard deviation to mean $\boldsymbol{B_1}$+ over the entire human brain ($\boldsymbol{B_1}+_{brain}$), or over the central transverse slice ($\boldsymbol{B_1}+_{slice}$)).

Results and discussion: Although the variation of the entire array performance (estimated as the ratio $\boldsymbol{B_1}+_{brain}/\sqrt{\boldsymbol{P}_{\mathbf{transmit}}}$) is more than 62% for all arrays analyzed, the ratio of $\boldsymbol{B_1}+_{brain}/\sqrt{\boldsymbol{P}_{\mathbf{brain}}}$ is equal to 1.0 µT/√W (for head #1) and 0.92 µT/√W (for head #2) with only +/-6% variation. When the conservative electrical field is smaller than the non-conservative electrical field, the array's $\boldsymbol{SAR_{10g}}$ related performance (the ratio $\boldsymbol{B_1}+_{brain}/\sqrt{\boldsymbol{SAR_{10g}}}$) is equal to 0.73 µT/√(W/kg) (for head #1) and 0.71 µT/√(W/kg) (for head #2) also with only +/-6% variation. In other words, each watt of power deposited in the brain corresponds to $\boldsymbol{SAR_{10g}}$=1.88 W/kg (for head #1) and $\boldsymbol{SAR_{10g}}$=1.68 W/kg (for head #2) with +/-15% variation. In such cases, the region of highest power deposition and SAR was found to be close to the top of the head (Figs.1,5 and 6). If the conservative electrical field becomes dominant the ratio $\boldsymbol{B_1}+_{brain}/\sqrt{\boldsymbol{SAR_{10g}}}$ can drop to 0.6 µT/√(W/kg) (for head #1) and 0.58µT/√(W/kg) (for head #2). The area of the highest power deposition and SAR was then found close to the skin of the nose (Figs.2,3,4).

For a head model symmetrically placed inside the coil, the conservative electrical field becomes dominant a) where the distance is shortest between an array element and the head model (well known), b) when the spacing between the array coil and its local shield becomes small, and c) when the tune/match/decoupling condition is suboptimal. Analysis of our large database shows that the sensitivity of $\boldsymbol{SAR_{10g}}$ in regard to the two latter conditions is smoothly varying and small (relative to the first condition). Step by step decrease of the distance between the array and its local shield results in a conservative electrical field that increases only rather slowly (Figs. 5, 6). Because the array current and conservative electrical field distributions depend on the tune/match/decouple condition, the ratio $\boldsymbol{B_1}+_{brain}/\sqrt{\boldsymbol{SAR_{10g}}}$ can vary by up to 9% for a given array geometry, if the conservative electrical field is dominant (Figs. 2,3). While the value of $\boldsymbol{B_1}+_{brain}/\sqrt{\boldsymbol{SAR_{10g}}}$ may remain stable, the profile of the 10-gram average SAR distribution may vary with array geometry and tuning/matching/decoupling. For example, increasing the array length results in the spread of a high SAR area deeper in the brain (Figs.1,5).

Conclusion: If the non-conservative RF electric field is generally dominant, even a significant variation of the array's current distribution due to changes in array geometry and different tuning/matching/decoupling has a very weak influence on the ratio $\boldsymbol{B_1}+_{brain}/\sqrt{\boldsymbol{SAR_{10g}}}$, although the SAR profile varies. As a result, the same MRI RF pulse sequence may produce nearly the same $\boldsymbol{SAR_{10g}}$ for different loop arrays, as long as similar $\boldsymbol{B_1}$+ maps are generated. This fact implies that optimization of loop array geometry and tune/match/decoupling conditions provides only a little room for improvement of array coil SAR performance. Instead, such optimization can be successfully used to significantly improve $\boldsymbol{B_1}+_{brain}/\sqrt{\boldsymbol{P}_{\mathbf{transmit}}}$ performance.

[1] A. Weisser, T. Lanz, Proc. ISMRM. 14 (2006) 2591. [2] M. Kozlov, R. Turner, Journal of Magnetic Resonance 200 (2009) 147–152.
[3] M. Kozlov , T. Rothe , R. Turner, Proc. ISMRM. 17 (2009) 4779.

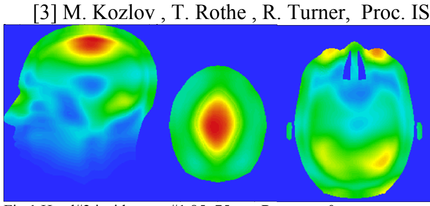
Fig.1 Head#2 inside array#1 85x75mm $\boldsymbol{P}_{\mathbf{array\_refl}}$=0

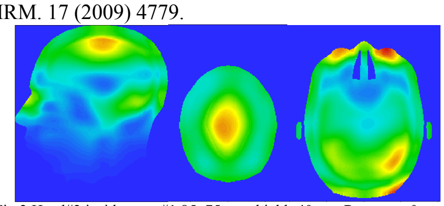
Fig.2 Head#2 inside array#1 85x75mm shield=40mm $\boldsymbol{P}_{\mathbf{array\_refl}}$=0

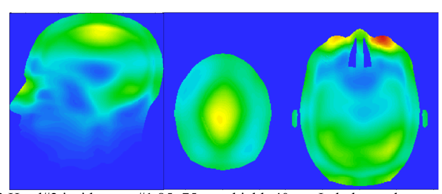
Fig.3 Head#2 inside array#1 85x75mm shield=40mm Ind. decoupl

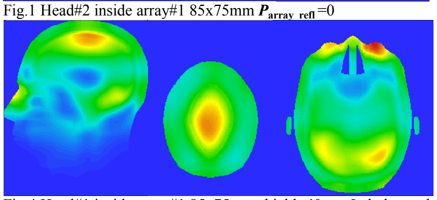
Fig.4 Head#1 inside array#1 85x75mm shield=40mm Ind. decoupl

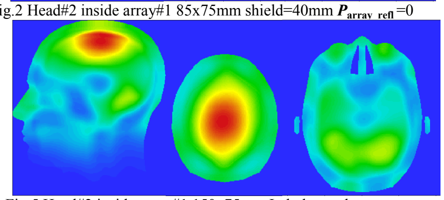
Fig.5 Head#2 inside array#1 150x75mm Ind. decoupl

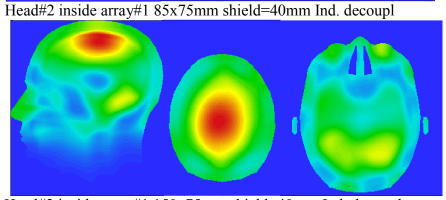
Fig.6 Head#2 inside array#1 150x75mm shield=40mm Ind. decoupl