A 16-Element Dual-row Transmit Coil Array for 3D RF Shimming at 9.4 T

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Introduction: For MRI at very high field strength, like 9.4 T, transmit array coils are an essential tool to mitigate B_1^+ field inhomogeneities caused by the shorter RF wavelength in tissue. Furthermore, it is challenging to create enough excitation in the cerebellum and lower temporal lobes. Additional flexibility to influence the B_1^+ field in the lower brain region can be achieved by unique coil layouts like the dual-row transmit element configuration [1, 2] which was numerically analyzed in [3]. However, most of these coils are transceiver arrays. In this work, we design an actively detunable 16-element dual-row transmit loop coil and investigate its static B_1 phase shimming performance.

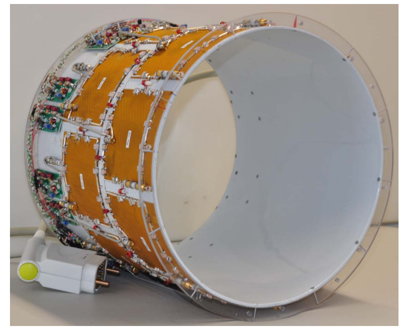

Fig. 1: 16-element, dual row transmit array with integrated TR switches and preamplifiers.

Methods: The transmit array was constructed on a 3 mm thick fiber glass cylinder with an inner diameter of 28 cm and 28.5 cm in length. Sixteen surface coils (2x8), 85 mm long and 100 mm wide, were arranged concentrically (Fig.1). The lower row was geometrically rotated by 22.5° compared to the top row. A gap of 12 mm was maintained between the elements on the same row and between the two rows. The neighboring elements on the same row and adjacent elements from the two rows were inductively decoupled. C-series capacitors were symmetrically distributed to tune the coil elements to 399.72 MHz; they were matched to 50 Ohms using a balanced matching network. Each coil element was connected to a home-built TR switch with preamplifier (WanTcom, MN, USA) assembled inside the coil housing (Fig.1). Experiments were performed on a 9.4 T MR scanner (Siemens Medical Solutions, Erlangen, Germany) equipped with a whole body magnet and a head gradient insert. The transmit RF power is split into 16 equal amplitude and phase components and connected to the coil element through the TR switch. For CP mode, a relative transmit phase delay of 22.5° on adjacent coil elements was realized, compensating for the additional cable length that was used for the lower row.

The phase shimming performance was investigated numerically using XFDTD 6.5 (Remcom). All loop elements in the coil model were tuned to the Larmor frequency before harmonic excitations were performed to get the steady-state RF field produced by each loop. The single-channel fields were scaled to 1W net input power and exported to MATLAB (The MathWorks). To find appropriate shim solutions, an optimization algorithm varied the phase offsets in order to minimize one of two cost functions: The inverse of the minimum B_1^+ field magnitude (CF1), in order to penalize field voids, or the B_1^+ standard deviation divided by the minimum B_1^+ value (CF2) in the corresponding region-of-interest (ROI). When an optimized phase setting was found, we calculated B_1^+ inhomogeneity inside the ROI (B_1^+ standard deviation divided by average B_1^+), as well as the maximum 10g SAR value that appeared in the model.

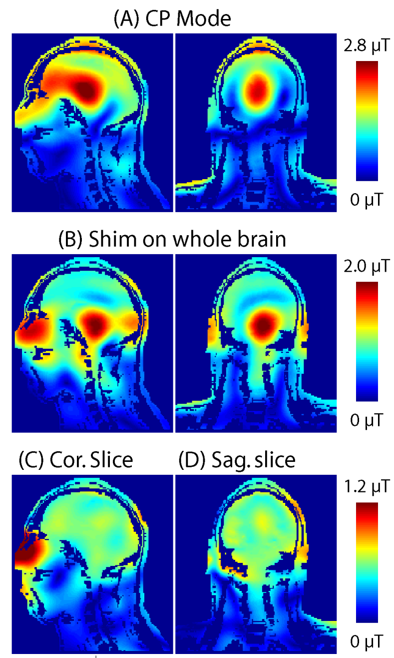

Fig. 2: B_1^+ field in CP mode (A), optimized for the whole brain (B) and for single slices (C and D). Note the different color scaling.

Results: Each coil element was tuned and matched to a head and shoulder phantom filled with tissue equivalent solution. The isolation between elements of adjacent rows was better than -20 dB and the isolation between the neighboring elements of the same row averages around -15 dB.

Figure 2A shows the B_1^+ field in the standard CP mode. Despite the large physical coverage of the array, there is still considerable inhomogeneity across the brain, especially in the temporal lobes and the cerebellum. A shim solution that improves homogeneity by more than 10% was found by optimizing for the whole brain using CF1 (Fig.2B). Note that no field voids appear in the temporal lobes any more. Figures 2C and 2D show shim solutions that aim for a homogeneous excitation (CF2) in the coronal and sagittal center-slices. An inhomogeneity of less than 10% could be achieved, but at the cost of elevated SAR problems due to the weaker average B_1^+ field, see Table 1. For *in vivo* validation, we performed a phase shimming experiment [4] in which we optimized for a single coronal slice using CF1 (Fig.3).

Conclusion: Initial results from an actively detunable dual-row transmit coil demonstrate lower brain coverage and the capability to mitigate transmit field inhomogeneity even in coronal and sagittal slices with a FOV covering the whole brain. This array will be combined with a tight fitting receive array for highly sensitive parallel signal reception in the future.

References: [1] Adriany G et al. Proc. ISMRM 2007 p166, [2] Avdievich NI et al. Proc. ISMRM 2011 p328, [3] Kozlov M et al. Proc. IEEE EMBS 2011 p547 – 553, [4] Hoffmann J et al Proc. ISMRM 2010 p1470

	Average B_1^+ in ROI [μT]	B_1 inhomogeneity in ROI [%]	Max. 10g SAR [W/kg]
CP Mode	1.34	38.44	13.53
ROI: Whole brain	0.96	27.14	9.06
ROI: Sagittal slice	0.50	9.07	9.56
ROI : Coronal slice	0.60	9.41	12.78

Table 1: Results for the presented shim solutions. Average B_1^+ and inhomogeneity were calculated only for voxels in the corresponding target ROI.

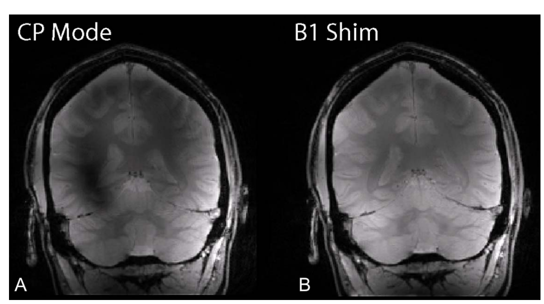

Fig. 3: Coronal FLASH acquired in CP mode (A) and with an optimized shim solution (B). Both images were acquired with the same transmit voltage and presented here with the same window level and without intensity correction.