A modular 16 ch. Transmit/32 ch. Receive Array for parallel Transmission and High Resolution fMRI at 7 Tesla

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Target Audience: RF Engineers, Researchers interested in pTX arrays

Purpose

The aim of this work was to develop a close fitting 7 Tesla fMRI whole head array capable of supporting both parallel RF transmission and receive array reception with the MR systems maximal available number of receive channels (32ch.). This is highly desirable since such a combined array both allows for full control of B_1^+(16 Ch.) and optimal receive performance [1-3]. An important second aim was to accommodate excellent task presentation capability and improve patient comfort.

Methods

A tight fitting coil former was built utilizing polycarbonate (PC) material for fused deposition molding FDM (Stratasys, Eden Prairie, MN, USA) (Fig. 1). The housing was designed to be modular and allows for a convenient swap between different sized transmit and receive coil combinations. Directly mounted to the lower holder are 24 receive only loop coils (Fig. 1B). A base holder than contained the transmitter array and related RF components (Fig. 1C). We improved our previous design [1] with the addition of a frontal cortex module containing 4 transmit coils and 8 receive only coils (Fig. 1A,D,E), this module fits snug with the main holder. With this addition we can achieve whole brain coverage (Fig 2.A). For greater performance using pTX multiband excitation and whole brain 3D B1+ control we utilized a dual row transmit array design both in the bottom and the top holder [4,5]. In the bottom holder this was realized with 2x6 coils of ~8x8cm^2 each. Coil decoupling was achieved with overlap decoupling between neighboring coils. An RF shield optimized for minimal eddy currents (Sheldahl, Northfield, MN) was placed 2.5 cm from the TX arrays for reduced interaction with the Magnet environment. Serial PIN diodes (M/A-com, Lowell, MA) in each TX coil loop were used for active transmitter detune. Receive coil elements varied in size from ~6cm to ~8cm and where built from 18 AWG(~2mm) copper wire, matched with a lattice balun. A major challenge was the positioning of receiver hardware in the limited space (~20mm) between transmitter and receivers. We solved this by utilizing a 1cm^2 small commercial low noise preamplifier board (WanTcom, Minneapolis, MN, USA) soldered to a customized~4cm^2 board with the required preamplifier protection, coil detune circuitry and phase shifters [6-8]. Maps for B_1^+ shimming were acquired using a fast, low flip angle estimation technique [9]. MRI experiments were conducted on a 7T scanner (Siemens Healthcare, Erlangen, Germany) equipped with 16x1kW power amplifiers and 32 receive channels.

Results and Discussion

Excellent decoupling in excess of -50dB between transmit array and detuned receive array was achievable and essential to ensure preamplifier protection. With this we succeeded in integrating 16TX +32RX coils into a very tight modular holder. The dual row transmit coil arrangement allowed for different shim solutions covering the entire brain: i) CP-like mode (see Fig.3A) .ii) applying a homogeneous phase shim solution could improve the transmit homogeneity quantified by the coefficient of variation (CV=std/mean) from CV=40% obtained with CP mode to 19.7% using the same RF power (see Fig3B). Further reduction towards CV=16% while using the same RF power can be achieved using magnitude+phase shimming algorithm. We plan to utilize this array primarily for advanced pTx applications such as pTX multiband pulses for functional or diffusion weighted imaging

Conclusion

Integration of 16 TX coils and 32 Rx coils into a tight fitting holder was established. The achievable homogeneity is significantly improved compared to a 1TX32RX whole brain coil (Nova Medical, Wilmington,MA,USA). The major advantage of the presented coil is increased pTX capability due to the dual row design and significantly improved task presentation.

References: 1. Wiggins, G. et al., Proc 17th ISMRM: #393,(2009) 2. Adriany, G. et al. Proc. 20th ISMRM , #429 (2012) 3. Zhao,W.et al. MRM **72**:291-300 (2014) 4. Adriany, G. et al. Proc. 15th ISMRM:#166 (2007) 5. Shajan G. et al. MRM **71** 870-879 (2014) 6. Keil, B.et al. MRM **66,** 1777-1787 (2011) 7. Reykowski, A., MRM. **33**(6): p. 848-52 (1995). 8. Roemer, P.B., et al., MRM**, 16,***2 192-225* (1990). 9. Van de Moortele, P.F., et al., Proc. 19th ISMRM:#367 (2011)

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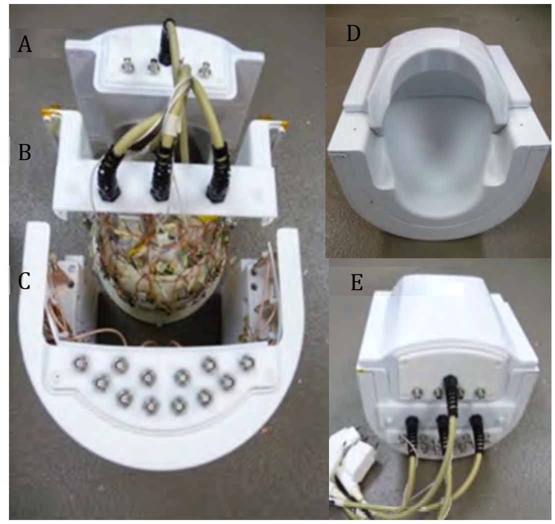

Fig. 1 Modular Holder Design. **A**: 4 Loop TX / 8 Receiver Top . **B**: Formfitting 24 Loop RX Holder **C**: 12 Loop TX, Dual Row Layout 6+6 **D,E:** Combined Holder

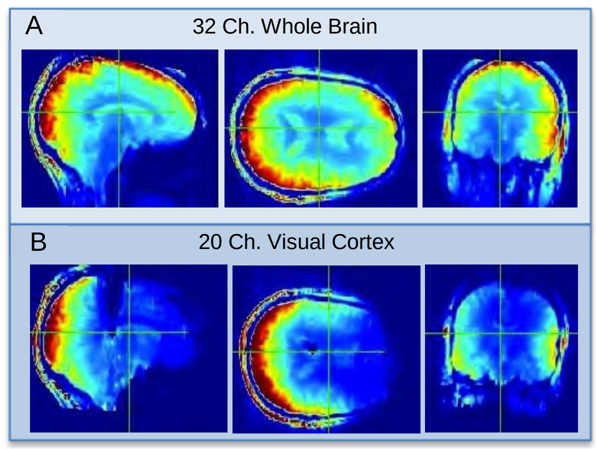

Fig. 2: SNR Comparison demonstrating increase coverage due to Top holder: **A:** 32 RX (Bottom Holder: 24 RX, 5 rows(4+4+6+6+4), Top: 8RX, 2 Rows (4+4). **B**: 20 RX, Bottom Holder only: 4 rows:4+6+6+4

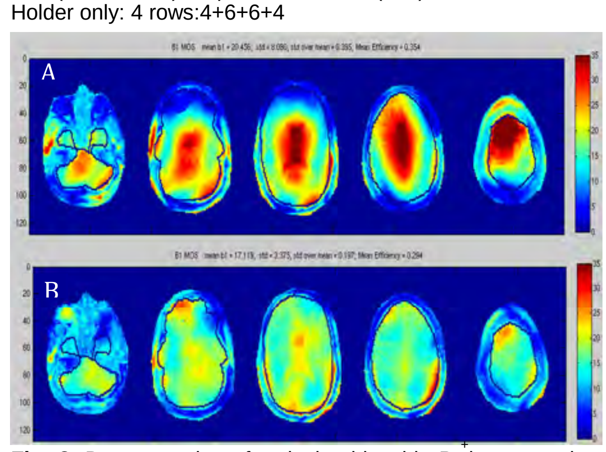

Fig. 3. Demonstration of typical achievable B_1^+ homogeneity improvements utilizing only phase and amplitude shimming (no pTX).**A** : CP Mode **B**: Phase/Magn.