# A 32 Channel Receive-only Head Coil And Detunable Transmit Birdcage Coil For 7 Tesla Brain Imaging

G. C. Wiggins[1], C. J. Wiggins[1], A. Potthast[2], V. Alagappan[1], O. Kraff[3], A. Reykowski[4], L. L. Wald[1]

[1]Athinoula A. Martinos Center for Biomedical Imaging, Massachusetts General Hospital, Charlestown, MA, United States, [2]Siemens Medical Solutions, Inc, Malvern, PA, United States, [3]Department of Diagnostic and Interventional Radiology, University Hospital Essen, Essen, Germany, [4]Invivo Corporation, Gainesville, FL, United States

**Introduction** The phased array [1] provides a particularly attractive solution for MR imaging at 7 Tesla. Increases in tissue loading at high frequency favor the use of arrays with many small receiver coils, and accelerated parallel reconstruction techniques can decrease scan times and reduce susceptibility distortions. There are however many challenges in their design and use at fields of 7 Tesla and higher, including coil to coil and cable interactions. Since body transmit coils are not routinely available on 7 Tesla scanner systems, it is also necessary to incorporate a detunable transmit coil. While many favor gapped array designs with the aim of improving g-factor in SENSE acceleration, recent analysis suggests that for large numbers of coils (<16) the SNR which can be achieved with accelerated imaging techniques is higher for overlapped coil designs [2]. In this study we extend the 32 channel soccer ball geometry previously described at 3T [3] to 7T with a hybrid birdcage transmit coil and extended measures for reducing cable interactions. This device was designed to work inside a 36cm diameter head gradient coil. This prototype was evaluated in human imaging for SNR and g-factor performance. .

**Methods** The system was developed and tested on a prototype 7T human scanner (Siemens Medical Solutions, Erlangen, Germany). The coils were arranged on a close-fitting fiberglass helmet modeled after the European head standard form EN960/1994 for protective headgear (222mm in AP, 181mm in RL, and 210mm in SI) (fig. 1). Circular surface coils were placed on the helmet, 26 with diameter of approximately 85mm and 6 with diameter of approximately 60 mm [3]. The coils were cut from Pyralux flexible circuit board material (Dupont) with a conductor width of 5 mm. Each coil had 4 or 5 gaps bridged with capacitors. A simple capacitive match and diode detuning circuit were incorporated into each coil. A cable trap was placed on each cable where is attached to the coil, formed by looping semirigid coax and bridging the loop with a capacitor, to block common mode currents. The coils were connected to the preamps with G02232 cables (Suhner) with cable lengths of 31.5cm and 22.5cm (not including the 5.4cm semirigid cable in the cable trap). The longer cable length is needed both to achieve preamp decoupling [1] and also to physically reach the coils lower on the helmet. The shorter cables were used in conjunction with a $\pi$-circuit phase shifter on the preamp board which provided sufficient phase length to allow preamp decoupling. Any further slack was removed by folding the cables back on themselves and tying them with cable ties to minimize the effective length projected to the transmit RF field. 32 preamps were installed on circuitboards mounted behind the receive and transmit coil assembly (fig 1), with circuitry to provide bias current to the PIN diode traps on the coils and a cable trap before each preamp input. The preamp outputs and PIN diode control lines were connected via two cable bundles directly into the scanner receiver electronics. Coupling between nearest neighbors was measured as the array was assembled by making an S12 measure transmitting on one coil and receiving on the coil next to it. Preamp decoupling was implemented by controlling cable length, use of the phase shifter, and looking for the decoupling minimum in an S12 measure with a coupled probe loosely coupled to the receive coil. A detunable hybrid birdcage coil was constructed with 28cm diameter and 16 rungs of 20cm length.

SNR maps were generated from proton density gradient echo images (TR/TE/flip/Slice = 200ms/4.07ms/20deg/3mm, 256x256, FOV = 220mm) obtained in human scans using the 32 channel phased array and also with the detunable transmit birdcage statically tuned with the detuned receive array in-place. SNR profiles were generated from the SNR maps. G-factor maps were calculated from the raw SNR scan data (including a noise image obtained with no RF excitation) by building the ratio between optimum SNR per pixel and calculated SENSE SNR per pixel. SENSE SNR is calculated by replacing the optimum weighting factors in the combination formula with the SENSE weighting factors.

**Results** The coupling between nearest neighbor coils on the receive array was -15dB on average, with the highest coupling being -11dB. The unloaded to loaded Q ratio for the receive coils was 8 and 5 for the larger and smaller coils respectively, indicating sample noise dominance. All the coils could have their preamp decoupling tuned with only minor adjustments to the input tank circuit on the preamp. Figure 2 shows SNR maps obtained with the receive array as well as with the transmit coil statically biased on and used in transmit/receive mode. The SNR maps (fig. 2) demonstrate SNR gains of 5 fold or more in the cortex and 2 fold in the center of the head when compared to the detunable transmit coil in T/R mode. The relative SNR performance can be seen more clearly in the SNR profiles in figure 3, taken through the center of the head in an axial scan. The higher SNR in the center of the head can be attributed in part to the small overall dimensions of the 32 channel helmet compared to the much larger transmit birdcage. Figure 4 shows maps of 1/g-factor, (allowing all the maps to be plotted with the same intensity scale). As expected the 7T 32 channel coil shows lower g-factors than a 7T 8 channel coil previously described [4], with maximum g-factors of 2.1 and 6.2 for rate 5 acceleration for the 32 and 8 channel coils respectively. The 7T 32 channel coil also shows lower g-factors than a 3T 32 channel coil [3] with an identical element geometry (g-max = 2.3 for rate 5), backing up previous theoretical predictions that g-factor improves with increasing field strength [5].

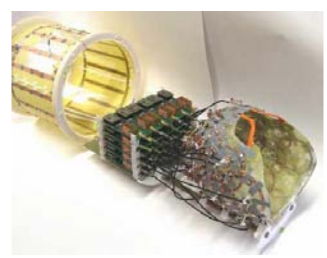

Figure 1: 32 channel receive array and preamps boards, removed from transmit coil for viewing.

**Conclusions** A 7 Tesla 32 channel phased array head coil and detunable birdcage transmit coil have been constructed and tested for human brain imaging. The phased array provides significant SNR improvements over the larger hybrid birdcage coil. Although the sensitivity is greatest near the coil elements, significant gains are observed throughout the brain, including in the center and at the vertex.

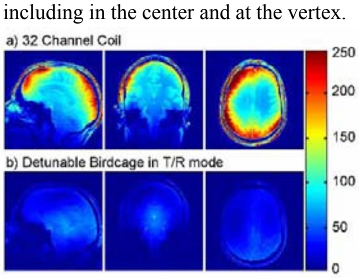


Figure 2: SNR maps for the 32 channel receive array and for the detunable transmit coil in T/R mode

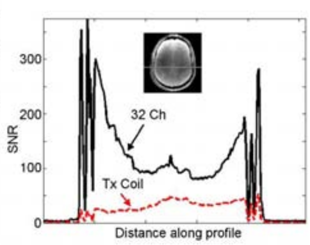


Figure 3: SNR profiles through an axial slice for the two coils. Gains of 5X or more in the cortex can be seen.

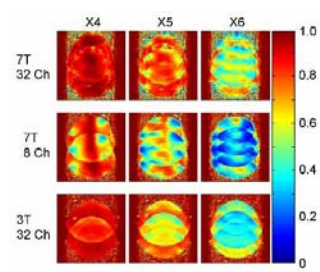


Figure 4: Maps of 1/g-factor demonstrating improved performance compared to 7T 8 Ch or 3T 32Ch

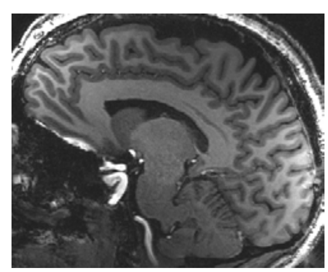

Figure 5: Un-normalized MPRAGE showing receive coverage. (0.6 x 0.6 x 1mm, TI/TR/TE/Flip = 900 / 2250/ 2.6 / 9, acquisition = 9min 46sec

[1] Roemer PB et al Magnetic Resonance in Medicine **16**, 192-225 (1990) [2] Wiesinger F et al Proc. ISMRM 2005 p672 [3] Wiggins et al Proc. ISMRM 2005 p671 [4] Wiggins GC et al Magn Reson Med 54(1):235-40 (2005) [5] Wiesinger F et al Magn Reson Med 52(2):376-90 (2003)

**Acknowledgements** Funding support from Office of National Drug Control Policy, Counterdrug Technology Assessment Center, NCRR Regional Resource grant P41RR14075 and the MIND Institute