A 96-channel MRI System with 23- and 90-channel Phase Array Head Coils at 1.5 Tesla

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Introduction A phased array of small surface coils placed close to the body provides increased SNR [1] and the option of using parallel acceleration techniques. As the number of available receive channels (n) increases, theoretical analysis predicts that it is possible to reduce the size of the individual elements, increasing SNR gains near the array, without losing sensitivity at larger distances; the increased number of coils making up for the poorer depth performance of the small individual coils [2]. Additionally, an increased number of channels is expected to enhance parallel image acceleration allowing higher factors without excessive gfactor penalty, and more flexibility of the choice of accelerated direction.

In order to explore the potential capabilities of large-n arrays for clinical applications, we constructed a 23 and 90 channel receive-only head coils and 96 channel RF receive capability for a 1.5 Tesla clinical scanner. The array coils utilize a novel geometry incorporating pentagonal and hexagonal symmetries which allow coils to be arranged over the dome of the head in a continuous overlapped array. The layout maintains the critical overlap between neighboring coils to minimize their mutual inductance. These prototypes were evaluated in phantom and human SNR, g-factor, noise correlation and imaging measurements.

Methods The system was developed and tested on a modified Avanto 32 channel 1.5T clinical scanner (Siemens Medical Solutions, Erlangen, Germany). This system was expanded to 96 channels by adding two more RF receiver racks with 32 receive channels each synchronized with the existing 32 receive channels. The RF signal from the coils were routed via low noise preamplifier/PIN diode bias assemblies mounted behind the coil. For the 23 channel coil, and for the first 32 channels of the 90 channel coil, the preamp outputs were connected to the plugs provided in the patient table. The additional 64 channels were routed to the additional amplifier stages and receivers. The digital output of the additional receiver racks were connected to two additional image reconstruction PCs synchronized to the main image reconstruction PC. The raw data of the MR signal that the three PCs collected during scans was reconstructed offline using either the commercial image reconstruction environment or home-developed software.

The individual receive 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). An arrangement of hexagonal and pentagonal tiles was created which covered the helmet, similar to a geodesic tiling of a sphere. Each tile had sides of approximately 44mm in the case of the 23 channel coil and 23mm for the 90 channel coil. With the 90 channel helmet it was necessary to distort the lattice around the pentagonal tiles in order to map it onto the surface of the helmet. Circular surface coils were placed on the helmets, each one centered on one of the tiles. To achieve the critical overlap, the coils corresponding to hexagonal and pentagonal tiles were approximately 50mm and 40mm respectively for the 90 channel helmet, and 95mm and 75mm for the 23 channel helmet. For the 90 channel helmet, the surface coils were machined from 0.031" inch thick G10 copper clad (1Oz/ft^2) circuit board with a conductor width of 2.5 mm. Patches of up to 27 overlapped circular coils were machined from single pieces of circuitboard (Fig. 2) before being attached to the helmet. The coil elements for the 23 channel coil were cut from Pyralux flexible circuit board material (Dupont). Each coil had 4 or 5 gaps bridged with capacitors, a detuning trap.

-SNR comparison were made with proton density gradient echo images (TR/TE/flip/Slice = 200ms/3.9ms/20deg/3mm, 256x256, FOV = 200mm) using the NEMA standard technique on a head shaped loading phantom. The dome coils were compared to the commercial CP headcoil. To assess parallel imaging performance, g-factor maps were calculated 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. The noise correlation matrix was measured from a noise image with no RF excitation.

Results The unloaded/loaded Q of the individual coil elements on the 90 channel helmet were 150/70 for the 50mm coils, 130/60 for the 40mm coils and higher for the larger coils of the 23 channel array. The noise correlation matrix (Fig. 2) and the uncombined images (not shown) demonstrate that good isolation was achieved between the individual coil elements, apart from a few coil combinations in the 90 channel figure. This increased coupling is likely due to a combination of one or more coils which did not have their preamp detuning optimally set and less than optimum coil overlap for these pairs.

For the 23 channel helmet, the phantom SNR maps demonstrate gains of 4 fold in the cortex compared to the CP coil 1.5 fold at the approximate location of the center of the corpus collosum. For the 90 channel helmet compared to the CP coil, SNR gain in the cortex was 8.4 fold, but lower SNR was seen at the collosum (0.9x that of the CP coil). G-factor calculations suggest significant improvements in 4 fold acceleration with the 90 channel coil compared with the 23 channel array (which itself shows gains over commercial 8 channel coils). Slice averaged gfactors for 4 fold y acceleration drop from 1.22 to 1.11.

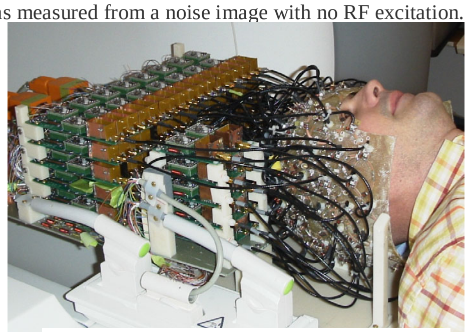
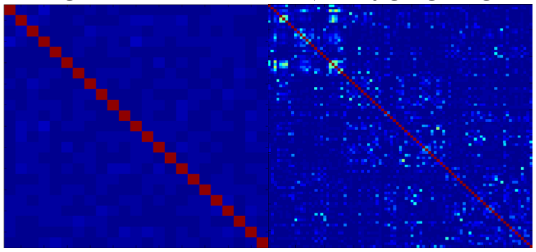


Fig.1 The 90 channel helmet array with preamps and biassin circuits.

Conclusions A 90 channel phased array coil and 96 channel receiver system have been constructed and tested in human brain imaging. These coils provide significant SNR improvements over commercially available head coils and allow for high parallel imaging accelerations. The higher SNR can be attributed in part to the close fitting design of the helmet coil former. Use of more but smaller coils improved cortical SNR more than central SNR. While the central SNR improved compared to the single channel volume coil for the 23 channel array, this was not the case for the 90 channel array. This deviation from the idealized theoretical calculations could be due to the relatively low unloaded to loaded Q ratio of these small coils. Maintaining tissue load dominance for these small coils might be facilitated by thicker conductors, lower loss cables, better RF fuses in the coils (seen to be a significant source of resistance) and by going to higher B0 field strengths.



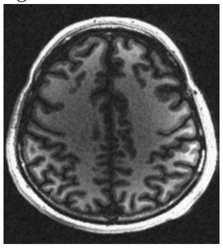


Fig 2 Photo of the 23 channel and 90 channel coil. Fig. 3 noise correlation matrix for the 23 and 90 channel arrays. Fig 4 MPRAGE from the 90 channel coil.

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