

A 32 Channel Receive-only Phased Array Head Coil for 3T with Novel Geodesic Tiling Geometry

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Introduction We describe a 32 channel receive-only 3T head coil with novel geometry incorporating pentagonal and hexagonal symmetries which allow coils to be arranged over the entire dome of the head in a continuous overlapped array. The layout maintains the critical overlap between neighboring coils to minimize their mutual inductance. A phased array of small surface coils placed close to the body provides increased SNR [1] and the option of using parallel acceleration techniques [2] [3]. As the number of available receive channels 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 individual coils [4]. This prototype was evaluated in phantom and human SNR, g-factor, noise correlation and imaging measurements.

Methods The system was developed and tested on a prototype 32 channel 3T clinical 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). 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 40mm. Circular surface coils were placed on the helmet, each one centered on one of the tiles. To achieve the critical overlap, the 26 coils corresponding to hexagonal tiles were approximately 80mm in diameter, while the 6 coils corresponding to pentagonal tiles were 60mm in diameter. 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 detuning trap, and were connected to the preamps with 370mm long G02232 cables (Suhner). 32 preamps were incorporated into the coil base, with circuitry to provide bias current to the PIN diode traps on the coils, a cable trap before each preamp input and a trimmer capacitor in series with the signal to tune preamp decoupling. The preamp outputs were connected via multistrand coaxial cable to plugs which fit into the sockets provided in the patient bed.

For SNR comparison, proton density gradient echo images (TR/TE/flip/Slice = 200ms/4.4ms/20deg/3mm, 256x256, FOV = 256mm) with identical parameters were obtained using the 32 channel phased array and also with a commercially available 8 channel head coil (MRIDC Gainesville FL). SNR maps were generated by dividing the smoothed image intensity by the standard deviation of the noise in a ROI outside the head. To assess parallel imaging performance, data were obtained with a WIP 3D FLASH sequence capable of acceleration in each of the 2 PE directions and an optimized sampling scheme of the calibration data required for the iPAT reconstruction (1mm isotropic, 256³ matrix, TR/TE/Flip = 12ms/ 4.76ms/15, BW 130Hz/pixel). An acceleration x12 was achieved (x4 for the in-plane phase encoding direction and x3 for the through-plane phase encoding direction), resulting in a total scan time of 1min 20sec. 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 were 283/24 for the 80mm coils and 210/32 for the 60mm coils. The noise correlation matrix (Fig. 2) and the uncombined images (not shown) demonstrate that good isolation was achieved between the individual coil elements. The SNR maps (Fig. 3) demonstrate SNR gains of 2 times or more in the cortex and 1.3 times in the center of the head when compared to the commercial 8 channel head coil. The higher SNR in the center of the head can be attributed in part to the close fitting design of the 32 channel helmet, which also resulted in higher image intensity inhomogeneity. G-factor simulations (Fig. 4) suggest that acceleration up to 4 x 4 = 16 fold may be possible with the use of small coils with g-factor related SNR loss limited to 40% max.

Conclusions A 3 Tesla 32 channel close fitting helmet phased array coil has been constructed and tested in human brain imaging. This coil provides significant SNR improvements over commercially available head coils, and allows for high parallel imaging accelerations to be used. Although the array sensitivity is highest in the cortex, significant improvements are observed throughout the brain, including the center of the head. A 12x accelerated 3D FLASH image (Fig. 5), obtained in 1:20, demonstrates the parallel imaging possibilities provided by the high SNR and large number of coil elements

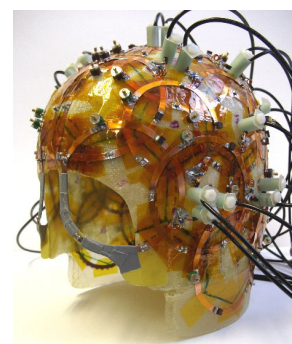


Figure 1. 32 channel phased array helmet

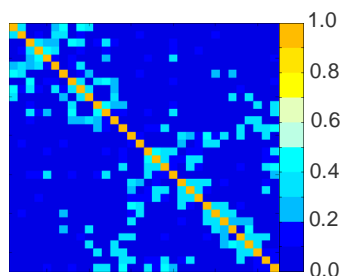


Figure 2. Noise correlation matrix for the 32 channel phased array. Maximum coil to coil noise correlation is about 40%

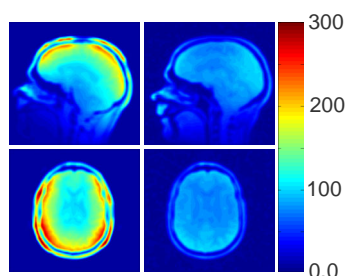


Figure 3. SNR maps for 32 channel helmet coil (left) and 8 channel commercial array coil.

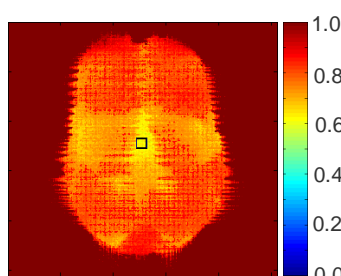


Figure 4. Simulated SNR loss with 16 times acceleration (4x AP, 4x RL), shown as 1/G-factor.

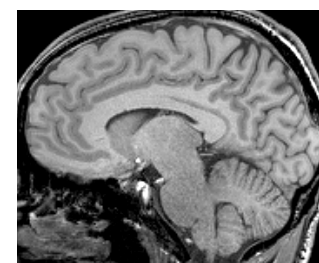


Figure 5. 3D GRE accelerated 12 fold (3x for in-plane PE direction, 4x for through-plane PE direction) acquired in 1:20 minutes. Voxel size = 1 x 1 x 1mm.

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