# A 32-channel Receive Array Coil for Pediatric Brain Imaging at 3T

**V. Alagappan[1,2], G. C. Wiggins[1], J. R. Polimeni[1], and L. L. Wald[1,3]**

[1]Department of Radiology, A.A Martinos Center for Biomedical Imaging, Charlestown, MA, United States, [2]Department of Biomedical engineering, Tufts University, Medford, MA, United States, [3]Health Science and Technology, MIT, Cambridge, MA, United States

**Introduction:** Pediatric MRI currently requires general anesthesia to preclude motion artifacts. With highly accelerated image encoding in conjunction motion navigators it may be possible to significantly reduce or even eliminate anesthesia. For research studies of brain development where anesthesia is not possible, improved motion robustness can significantly improve the success rate. Highly parallel array detection is thus critically needed for pediatric brain imaging. In this work we have designed and constructed a 32-channel receive-only head coil to image 4 year olds. The sensitivity and image acceleration achievable with an optimized 32 channel pediatric brain array is compared with a commercially available 32-channel adult head coil and a 12-channel adult head coil.

**Methods:** The array was developed and tested on a 32-channel 3T clinical scanner (MAGNETOM Trio, A Tim System; Siemens Healthcare, Erlangen, Germany). The coils were arranged on a close-fitting fiberglass helmet scaled down from an adult (95th percentile) helmet to the relative head circumference for 4 yr olds. The helmet dimensions were 17.5cm AP, 16.8cm RL. The layout of the coil elements consisted of an arrangement of hexagonal and pentagonal tiles, similar to a truncated icosahedral (soccer ball) tiling of a sphere(1). The entire shape incorporated 27 hexagons and 5 pentagons. Wire coil elements (18awg tinned copper wire) with a diameter of 65mm (hexagonal tile positions) and 52mm (pentagonal tile positions) were placed at the location of the pentagons and hexagons. Neighboring coils were critically overlapped to reduce coupling. Each loop incorporated two capacitors. A detuning circuit (2) was formed across the match capacitor using a variable inductor (Coil Craft, Cary, IL) and a PIN diode (Macom, MA4P4002B-402, Lowell MA). This active trap circuit provided the primary detuning of the receive element during transmit. Fuses with a rating of 570 mA where added as an additional protection mechanism if the detuning trap circuit fails. The preamplifiers (Siemens Healthcare, Erlangen Germany) were placed adjacent to the coil (after 4.5cm coax.) Careful control of the cable length and tuning of the preamplifier input matching circuit provided preamplifier decoupling.

For SNR comparison, proton density weighted gradient echo images (TR/TE/flip = 300ms/6ms/10°, slice=5mm, 384×384, FOV=192mm, BW=300 Hz/Pixel) were obtained using a child's head shaped water phantoms. Noise covariance information was acquired from the same pulse sequence but with no RF excitation. The SNR maps are calculated for an optimal SNR reconstruction (incorporating noise covariance information) (3). The G-factor maps were also calculated for the phantom data. The coil was compared to the commercially available Siemens 32-channel adult 3T head coil built with similar soccer ball design and the Siemens 12-channel adult head coil. Finally the array performance was tested in high resolution anatomical imaging.

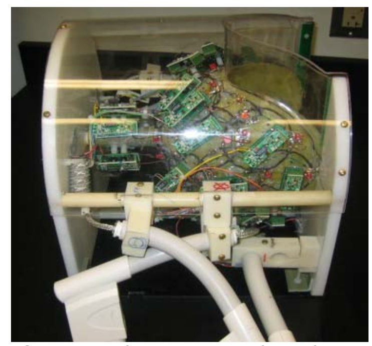

**Figure 1** : The constructed 32-channel pediatric array coil .

**Results:** The array showed typical $S_{12}$ decoupling between nearest neighbors of -15 dB and preamplifier decoupling of -25 dB. Typical unloaded-to-loaded Q values for the elements was 4.2 ($Q_{UL}/Q_L$ = 252/60). Figure 2 shows the SNR comparison for all the three coils in all the three imaging axes. Average SNR over the three slice planes was increased 1.5 fold compared to the 32ch adult array and 2.7 fold compared to the adult 12ch array. Near the edge of the phantom the SNR gains were 2 and 4 fold compared to the two adult arrays. In the center an SNR increase of 1.2 and 2 fold were measured. Figure 3 shows the 2D 1/G-factor maps for different acceleration rates on all the three coils along with the maximum G-factor. Maximum G factors were reduced significantly (e.g. 1.7 fold for 2x3 fold acceleration). Figure 4 shows a sagittal slice in a high resolution (0.75mm isotropic) 3D MPRAGE image acquired in under 4 minutes with 3x GRAPPA acceleration.

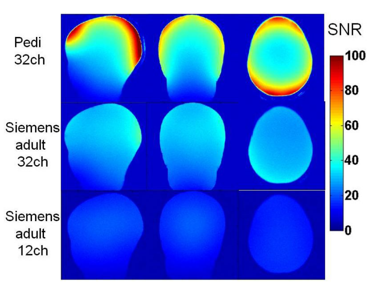


**Figure 2** SNR maps for a head shaped phantom.

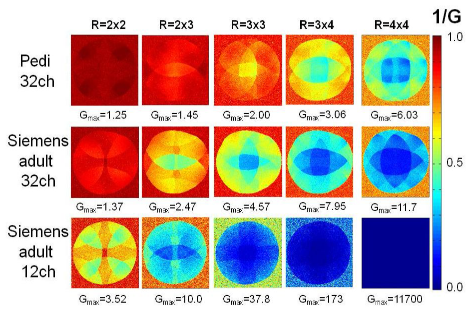


**Figure 3** 2D 1/G-factor maps on an axial slice for the 3 arrays.

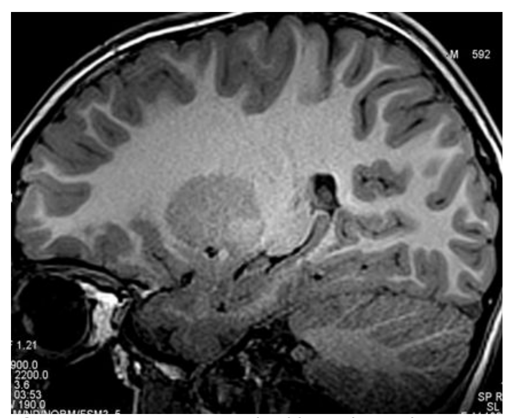

**Figure 4** MPRAGE acquired in under 4min at 750um isotropic spatial resolution (R=3 GRAPPA), TI/TR/TE=900/220/3.6ms

**Conclusions:** The 3T, 32-channel phased array head coil optimized for pediatric imaging has been constructed and tested *in vivo*. The significant SNR and G-factor performance suggests that 1) there is a significant penalty for using an incorrectly sized head array and 2) the smaller head-size of children likely makes close-fitting arrays of small coils particularly favorable for pediatric brain imaging.

**References:** [1] Wiggins et al. Magn Reson Med 2006;56(1):216-223. [2] Edelstein et al. JMR 1986 67 p156-161. [3] Kellman et al. Magn Reson Med 2005; 54(6): 1439-1447.