Age-Optimized 32-Channel Brain Arrays for 3T Pediatric Imaging

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Introduction: Compromising the size and shape of pediatric brain arrays so that "one size fits all" or using adult brain or knee arrays causes a significant degradation of SNR and parallel imaging performance compared to a coil of the appropriate size and shape for a given aged child [1]. Unfortunately, rapid head growth in the first years of life requires either a flexible array approach or multiple sizes which span the size range with reasonable discrete increments. In this work, we developed and tested four incremental sized 32-channel receive only head coils for pediatric patients spanning an age range of 6 months to 7 years old. We based the increments on brain growth charts and average head shapes taken from clinical MRI studies. This led to the construction of arrays nominally sized for 6mo, 1yro, 4yro and 7yro. The sensitivity and image acceleration achievable using these optimized 32-channel pediatric brain arrays is compared with a commercially available 32-channel adult head array and a pediatric sized single channel birdcage. The tailored array approach underscores the inefficiency of using adult or single channel coils for pediatric brain imaging showing significant SNR gains for both accelerated and unaccelerated imaging.

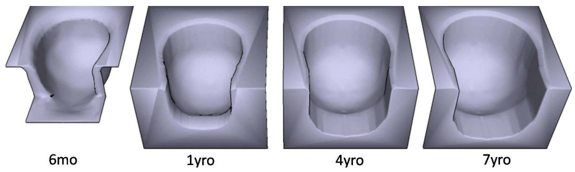

Fig. 1. 3D models of all the age-matched helmets for the respective averaged children head shapes.

Material and Methods: The coil consists of two parts, a deep posterior segment covering all but the face and forehead designed so the child can lie down into the coil (rather than a helmet which comes down over the head) (Fig. 1), and a separate "frontal paddle" over the forehead (Fig. 2). The eyes and face are completely unobstructed to facilitate anesthesia or functional studies. The different helmet sizes were obtained from the surface contours of aligned 3D MRI scans from N~20 corresponding aged children. The helmet shape was taken by dilating the 95% contour to accommodate 3mm foam padding. The final design of the helmet parts were printed in ABS plastic using a rapid prototyping 3D printer (Dimension, Inc., MN, USA).

The layout of the wire (16awg) circular coil elements consisted of a hexagonal and pentagonal tiling pattern [2]. The shape of the bottom part incorporates 23 hexagons and 5 pentagons. The top part completes the 32-channel with additional 4 overlapped oval elements. Neighboring coils are critically overlapped to reduce coupling. Each loop incorporates two capacitors, with a conventional matching with the coax cable across one capacitor. An active detuning circuit was formed across the match capacitor using a variable inductor (Coil Craft, Cary, IL, USA) and a PIN diode (Macom, MA4P4002B-402, Lowell MA, USA). In addition to this primary detuning during Tx, series fuses with a rating of 570 mA where added to each loop as additional protection during transmit. All elements were tuned to 123.25 MHz and matched a load impedance designed to minimize the noise figure of the preamplifiers (Siemens Healthcare, Erlangen Germany). The input impedance of the preamplifiers (Siemens Healthcare, Erlangen Germany) was transformed by the 5.4cm cable connecting it to the coil to provide a low impedance across the PIN diode and thus a high impedance in the loop (preamp decoupling). The preamps where orientated to minimize Hall effect issues [3].

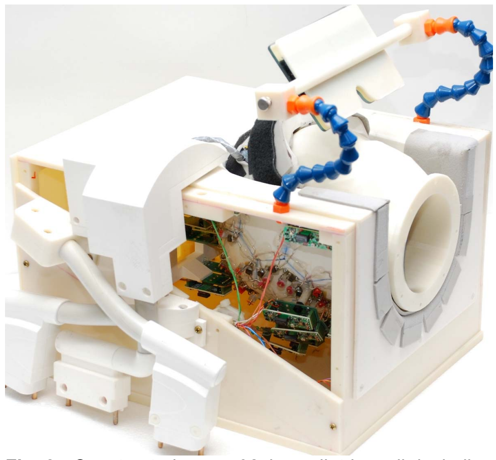

Fig. 2. Constructed 4yro 32ch pediatric coil including paddle, mirror, and head shaped phantom.

Data were acquired on a 3T clinical whole body MR Siemens scanner (MAGNETOM, Trio a Tim system, Siemens Healthcare, Erlangen Germany). For SNR and G-factor comparison, proton density weighted gradient echo images (TR/TE/flip=30ms/6ms/30°, slice=6mm, 192×192, FOV=170mm, BW=200 Hz/Pixel) were obtained using 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) [4]. The coil was compared to the commercially available Siemens 32-channel adult 3T head coil built with similar soccer ball design and the Siemens extremity coil. Finally the array performance was tested in highly accelerated anatomical imaging using a non sedated child.

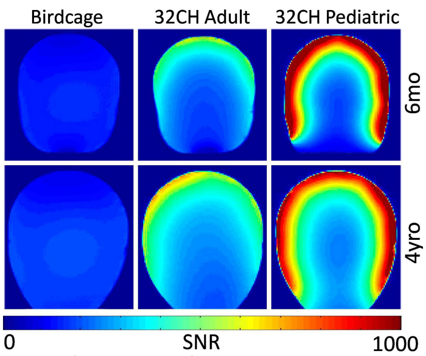

Fig. 3. SNR maps of a size matched phantom corresponding to a 6mo child and 4yro child in a CP birdcage knee coil, 32ch adult coil, and constructed pediatric 32ch coil.

Results: All four arrays show S_{12} decoupling between nearest neighbors of approximately -16dB and preamplifier decoupling of -25 dB. Typical unloaded-to-loaded Q values ranges from 3.9 for the 6mo coil to 8.9 for the 7yro coil. Fig. 3 shows a representative sample of SNR comparison of the developed 6mo and 4yro coil with the 32-channel adult coil and a knee birdcage coil. Near the edge of the phantom the SNR gains range from 5.1 fold for the 6mo coil and 4.3 fold for the 7yr old coil compared to the 32-channel adult coil. In the center, the SNR ranges from 1.2 and 1.7 fold. Averaged and Maximum G-factors were reduced significantly (e.g. 2.2 fold for 3x3 fold acceleration using 4yro coil instead of a 32-channel adult array) (Fig. 4).

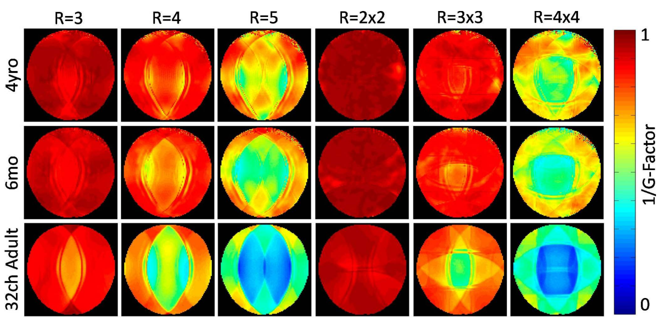

Fig. 4. Inverse G-Factor maps for different accelerations using age matched phantoms in comparison to a 32ch adult coil.

Conclusion: Four age-matched 32-channel phased-array head coils optimized for pediatric imaging have been developed and tested in vivo. The coils are well-suited for highly accelerated pediatric brain imaging and provide significant improvements in SNR. Time reduction due to highly parallel array detection improves motion related artifacts and avoids the usage of anesthetized children during MRI scanning.

Acknowledgment: P41 (P41RR14075) **References:** [1] V.Alagapan et al. Proc. ISMRM 2009 p.108, [2] G.C. Wiggins et al. MRM 2006, 56(1):216-23, [3] C. Possanzi et al. Proc. ISMRM 2008, P.1123, [4] Kellmann et al. MRM 2005 54(6):1439-1447.