Validation of A Semi-flexible 64-channel Receive-only Phased Array For Pediatric Body MRI at 3T

Tao Zhang[1,2], Joseph Y Cheng[1,2], Paul D Calderon[1], Thomas Grafendorfer[3], Greig Scott[2], Bob Rainey[3], Mark Giancola[3], Fraser Robb[3], John M Pauly[2], Brian A Hargreaves[1], and Shreyas S Vasanawala[1]

[1]*Radiology, Stanford University, Stanford, CA, United States,* [2]*Electrical Engineering, Stanford University, Stanford, CA, United States,* [3]*GE Healthcare, WI, United States*

Purpose: Pediatric MRI is often performed using receive coils designed for adult patients. The mismatch of the receive coils and the pediatric patients can cause degraded image quality [1]. In this work, a 64-channel receive-only phased array dedicated for pediatric body MRI is designed and constructed. This coil array incorporates the quarter-wavelength-balun technique, achieves ideal coil decoupling, and conforms to different pediatric patient sizes. Here, we validated the performance and flexibility of the coil array in phantom and *in vivo* studies.

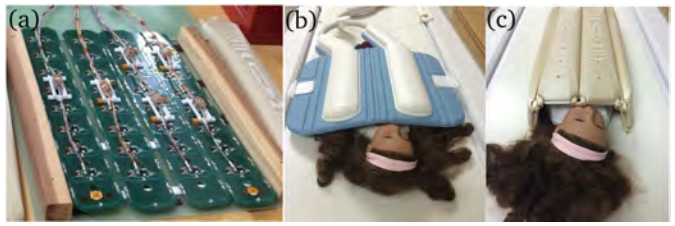

Fig 1. (a) Anterior 32 elements of the flexible coil before packaging: individual elements only overlap in the S/I direction, yielding high flexibility in the non-overlapping direction. Comparison of the anterior elements from a 32ch cardiac coil (b) and the 64ch pediatric coil (c) on a pediatric patient model demonstrates the flexibility of the 64ch pediatric coil.

Methods: The 64ch phased array (Fig 1) contains a 32ch anterior coil and a 32ch posterior coil embedded in the patient table. The quarter-wavelength-balun technique is used to replace individual baluns [2]. Individual coils only overlap in the S/I direction, which creates high flexibility in the non-overlapping (transverse) direction. To verify the coil decoupling and flexibility of the designed coil array, three different foam supports (Fig 2a) were designed to achieve 10°, 20° and 30° bending of the 32ch anterior coil. This simulates different matching scenarios of the receive coils and patient sizes. A 9”-diameter saline phantom was imaged on a GE 3T MR750 scanner using the 32ch anterior coil placed on the foam supports. For each bending scenario, a 2D spin echo sequence was performed. Normalized noise correlation matrix of the anterior 32 elements was calculated [3]. The datasets were retrospectively undersampled by factors of 2, 3, 4, and 5 in the phase encoding direction, and the corresponding g-factor was calculated. With IRB approval, two *in vivo* studies were performed. The first study was performed on a 30-year-old subject with breath-holding using a RF-spoiled gradient echo sequence with both the 64ch pediatric coil and product 32ch cardiac coil. The second study was performed on an 18-month-old subject free-breathing using the upper 32 elements of the 64ch pediatric coil.

Results: The sum of squares images are shown in Fig 2a. As the coil gets closer to the phantom, higher SNR was observed in the image. This demonstrates the advantages of flexible coils conforming to patient size and shape. The noise correlation matrices are shown in Fig 2a. The max and mean value of the noise correlation did not vary much in different scenarios, which demonstrates good coil decoupling. The g-factor maps with different acceleration factors are shown in Fig 2b. As the anterior peripheral elements tilted more towards the phantom, the g-factor decreased, especially under higher acceleration in the lower part of the phantom, where coil sensitivity variations were small. This also suggests that better parallel imaging performance can be expected when the coil array matches the patient size. Images from *in vivo* studies are shown in Fig 3. Similar image quality was observed for the first study using two different coils. Excellent image quality was achieved for the second study using the 64ch pediatric coil.

Conclusion: In this work, we have developed and validated a flexible 64-channel receive-only phased array for pediatric body MRI at 3T.

References: [1] Vasanawala S, et al. ISMRM 2011, p161. [2]. Grafendorfer T, et al. ISMRM 2014, p1319. [3] Schmitt M. Magn Reson Med 2008; 59: 1431-1439.

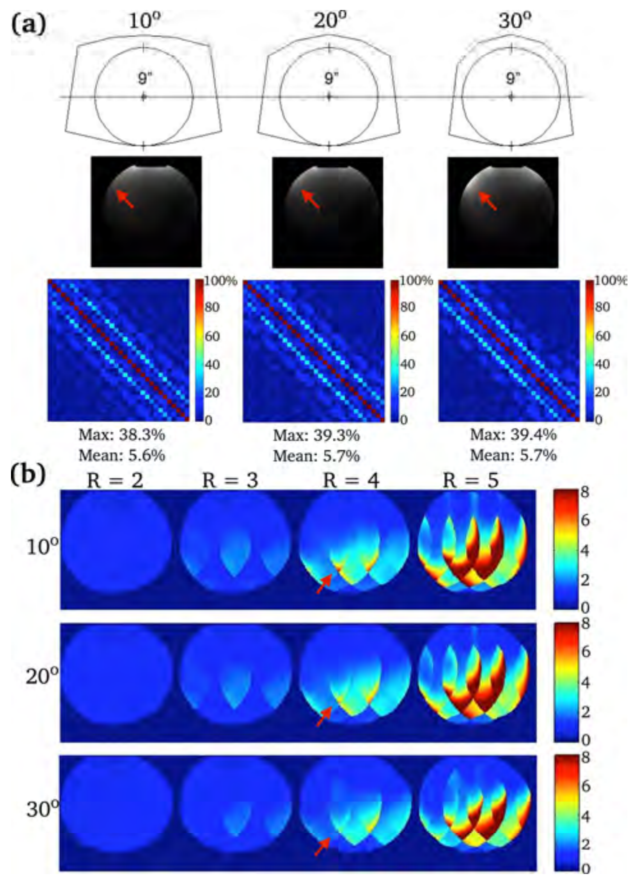

Fig 2. (a) Top: schematics of the form supports ensuring coil bending to 10°, 20°, and 30°. Middle: acquired images in three scenarios. Note that all images were displayed with the same window width/level. Higher SNR was achieved as the coil positioned closer to the phantom (arrows). Bottom: the corresponding noise correlation matrices. The max and mean value of the noise correlation matrices are also listed. The small noise correlation showed that good coil decoupling was achieved by the coil array. Also, noise correlation did not vary much in different scenarios. (b) g-factor maps for acceleration factors 2, 3, 4, and 5: as the anterior coil is bent further, the g-factor slightly decreased (arrows), which suggests better parallel imaging performance as the coil conforms to the subject.

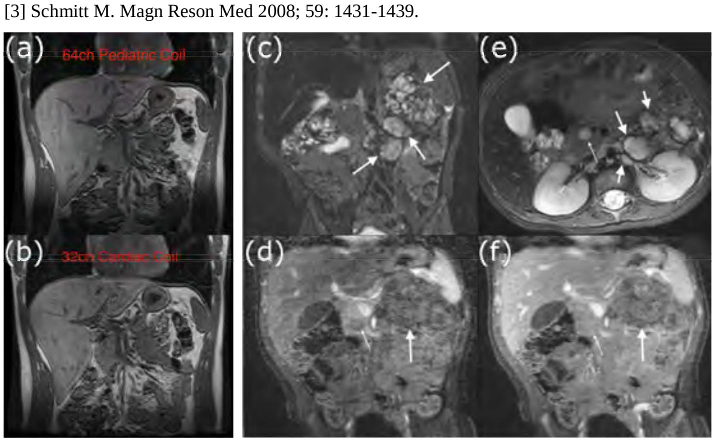

Fig 3. (a) Adult volunteer using middle 32 elements of the 64ch pediatric coil and (b) product 32ch cardiac coil had similar image quality. (c) An 18-month-old child imaged with upper 32 elements of the 64ch pediatric coil. Coronal T2 image shows good delineation of tumor (arrows), which is also seen on axial T2 (e), post-contrast arterial phase (d) and venous phase (f).