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

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Abstract

Purpose—To design, construct, and validate a semi-flexible 64-channel receive-only phased array for pediatric body MRI at 3T.

Methods—A 64-channel receive-only phased array was developed and constructed. The designed flexible coil can easily conform to different patient sizes with non-overlapping coil elements in the transverse plane. It can cover a field of view of up to $44 \times 28 \text{ cm}^2$ and removes the need for coil repositioning for body MRI patients with multiple clinical concerns. The 64-channel coil was compared with a 32-channel standard coil for signal-to-noise ratio (SNR) and parallel imaging performances on different phantoms. With IRB approval and informed consent/assent, the designed coil was validated on 21 consecutive pediatric patients.

Results—The pediatric coil provided higher SNR than the standard coil on different phantoms, with the averaged SNR gain at least 23% over a depth of 7 cm along the cross-section of phantoms. It also achieved better parallel imaging performance under moderate acceleration factors. Good image quality (average score 4.6 out of 5) was achieved using the developed pediatric coil in the clinical studies.

Conclusion—A 64-channel semi-flexible receive-only phased array has been developed and validated to facilitate high quality pediatric body MRI at 3T.

Keywords

MR phased array; pediatric MRI; parallel imaging; flexible coils; body coils

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Introduction

The lack of ionizing radiation makes MRI a very compelling imaging modality for pediatric patients. However, pediatric MRI is particularly challenging due to the small size of the patients, long scan time, and the need for general anesthesia (1–7). Parallel imaging (8–11), using phased arrays with multiple coil elements to acquire signal simultaneously from reduced field of view (FOV) acquisitions, has significantly reduced scan time over more than a decade. It has been successfully applied to pediatric MRI in several clinical applications (12–16).

Coil arrays with small coil elements and high channel counts can provide higher intrinsic signal-to-noise ratio (SNR), enable large volumetric coverage, and achieve better parallel imaging performance for higher acceleration than conventional coils (17–21). In recent years, various dedicated dense coil arrays have been developed for different imaging needs (22–28). Better image quality and parallel imaging performance have also been validated clinically (29–32).

Despite the active research and development in dense coil arrays, few commercially available coils dedicated to pediatric body MRI have been developed, and there is a need for a high density, high channel count pediatric coil array. In clinical practices, pediatric body MRI is often performed using receive coils designed for adults. The mismatch of the receive coils and the pediatric patients can compromise the parallel imaging performance and degrade image quality (1,12,33). Flexible coils have the potential to further improve image quality (23,34). Although some commercially available coils are flexible to conform to different adult patient sizes, the flexibility of the coils is insufficient for pediatric patients.

In this work, a 64-channel semi-flexible receive-only phased array dedicated for pediatric body MRI at 3T has been designed and constructed. This coil array features $\lambda/4$ Baluns (35, 36) and achieves excellent channel decoupling. The coil elements only overlap in the superior/inferior (S/I) direction, creating high flexibility in the transverse plane to conform to different pediatric patient sizes. Large coverage of the coil can also reduce the preparation time for coil repositioning for patients with multiple clinical concerns that require different MRI exams. The performance and flexibility of the coil array is assessed in phantom and through in vivo studies.

Methods

Phased array design and construction

A 64-channel pediatric body coil consisting of a semi-flexible 32-channel anterior coil installed in a flexible mechanical package and a rigid 32-channel posterior coil embedded in the rigid patient table is designed and constructed. Both anterior and posterior coil elements are arranged in an 8×4 array. The details of the coil array layout are shown in Fig. 1(a). Individual coil elements are kept small ($6.35 \times 6.35 \text{ cm}^2$) to achieve good central SNR and parallel imaging performance for children up to 8 years old (19, 21, 22). An equivalent schematic of a single coil element is illustrated in Fig. 1(b). The coil can cover approximately a $44 \times 28 \text{ cm}^2$ FOV, ideal for torso coverage of small pediatric patients. Coil

elements only overlap in the S/I direction, which creates high flexibility in the non-overlapping transverse directions. The distance between the centers of two adjacent non-overlapping coil elements in the right/left (R/L) direction is 7.24 cm. The constructed coil before packaging is shown in Fig. 1(c). The anterior mechanical package consists of three flexible hinge axes, preserving the flexibility of the anterior coil. As shown in Fig. 1(d), compared to a commercially available 32-channel standard flexible cardiac coil, the designed anterior coil has higher flexibility and can better conform to different patient sizes.

The unloaded to loaded Q ratio was approximately 6 for a single isolated coil element (the distance between the coil and the load (human abdomen) approximately 1 cm). This agrees with the value of a similar sized coil element reported in (17). Contrary to the conventional approach of placing individual Baluns for each channel, $\lambda/4$ Baluns (35,36) were employed in this work. The $\lambda/4$ Baluns approach is more deterministic and robust for channel decoupling (Fig. 2(a)). Keeping the ground connection at an exact location of $\lambda/4$ is not necessary for good channel decoupling performance. In this work, the ground connections for each row are set together at approximately $\lambda/4$ from the most superior coil element along the cable track to keep ground loop resonance modes away from the Larmor frequency. The cables are routed right through the middle of the coil elements for minimal coil-cable interactions. Floating cable traps were also integrated in the design (Fig. 1(c)) (37). Coil tuning and testing was performed on a human subject. Preamplifier (GE Healthcare, Waukesha, WI, USA) decoupling was approximately 100 ohms, and active decoupling was greater than 2000 ohms. Matching S11 was lower than -12 dB. Preamplifier decoupling was leveraged to ensure diagonal and next-nearest neighbor elements were suitably decoupled.

Due to the system limitation and different imaging needs, nine different coil modes were designed and demonstrated in Supporting Figure S1: (1) upper torso mode; (2) middle torso mode; (3) lower torso mode; (4) anterior mode; (5) posterior mode; (6) left hip mode; (7) right shoulder mode; (8) left shoulder mode; (9) right hip mode. Although all 64-channel coils were plugged into the system at the same time, the switching matrix can only support up to 32 coil elements.

Phantom studies

All the following phantom and in vivo studies were performed on a 3T MR750 scanner (GE Healthcare, Waukesha, WI, USA).

Channel decoupling—To validate the channel decoupling, the 32-channel flexible anterior coil was first placed on top of two unloaded rectangular phantoms with silicone oil (GE Healthcare, Waukesha, WI, USA). 2D gradient echo images were acquired in the coronal plane with the following acquisition parameters: matrix 256×256 , slice thickness 5 mm, TR 250 ms, TE minimum full, and flip angle 90° . Images from individual coil elements were compared to qualitatively assess the channel decoupling. The noise correlation matrix of the anterior 32 elements was calculated for a noise-only dataset acquired with similar acquisition parameters.

SNR comparison—To evaluate the SNR gain by the 64-channel flexible coil compared to a 32-channel standard coil (In Vivo Corporation, Gainesville, FL, USA), three different

types of phantoms were used: (1) unloaded rectangular phantom with silicone oil, (2) loaded rectangular phantom with 3.3685 g/L $\text{NiCl}_2 \cdot 6\text{H}_2\text{O}$ and 2.4 g/L NaCl (USA Instruments, Aurora, OH, USA), and (3) loaded pediatric torso phantom with 3.37 g/L $\text{NiCl}_2 \cdot 6\text{H}_2\text{O}$ and 2.5 g/L NaCl. This solution is within the concentration range recommended by (38) for loading (approximate conductivity 0.5 S/m and relative permittivity 70) and relative short relaxation times, and has been used in other works (22). For simplicity, only the anterior coils were compared. 2D axial spin echo images were acquired for all phantoms with the following acquisition parameters: TE 20 ms, TR 500 ms, slice thickness 10 mm, slice spacing 10 mm, and 9 slices in total. FOV was $36.0 \times 25.2 \text{ cm}^2$ for the unloaded rectangular phantom, $40.0 \times 28.0 \text{ cm}^2$ for the loaded rectangular phantom, and $20 \times 20 \text{ cm}^2$ for the loaded pediatric torso phantom. The corresponding matrix sizes were 256×180 , 256×180 , and 192×192 respectively. The same acquisition was performed twice for each phantom, the second time without radiofrequency pulses for noise acquisition. Coil sensitivities were estimated using ESPIRiT (39) from the first acquisition. Pseudo-multiple replica method was used for SNR calculation (40). Noise statistics were estimated from the second acquisition. New noise was generated based on the estimated noise covariance matrix and added to the k-space data from the first acquisition. A full image was generated using a SENSE reconstruction with $R = 1$ (10). The synthesis procedure was repeated 100 times for the pseudo-multiple replica SNR calculation. SNR maps and SNR profiles at selected locations were compared between the two coils for each phantom.

Parallel imaging performance—To evaluate the parallel imaging performance, the pediatric torso phantom was scanned again with both anterior and posterior coils for the 64-channel flexible coil and 32-channel standard coil (Fig. 4(a)). Lower torso mode was selected for the 64-channel coil. Similar to the previous experiment, fully sampled 2D axial spin echo images were acquired with the same acquisition parameters. Coil sensitivities were estimated using ESPIRiT, and a noise-only dataset was acquired to estimate the noise statistics. The dataset was retrospectively undersampled in two directions by different reduction factors: 2, 3, and 4 in the R/L direction, and 3 and 4 in the anterior/posterior (A/P) direction. The corresponding g-factor maps were calculated and compared between the two coils. Note that the prescribed FOV in the A/P direction was bigger than the phantom size, so the effective reduction factor in the A/P direction was approximately 1.5 and 2.

In vivo studies

With institutional review board approval and informed patient consent/assent, 21 consecutive patients (11 males, 9 females, and 1 gender undetermined) referred for body MRI at our institution were recruited from November 2014 to February 2015. The patient ages range from 1 day to 10.4 years (mean, 2.9 years). Patient demographics and clinical indications are summarized in Supporting Table S1. Four subjects required a combined spine and abdomen/pelvis exam. Therefore, the total number of exams in the study was 25. All imaging was performed using the designed 64-channel pediatric coil. Different coil modes were selected to cover the area of interest. Institutional standard clinical protocols were performed for each exam.

All patient exams were reviewed in consensus by two radiologists on a five-point scale. The scoring criteria were: non-diagnostic image quality (score = 1); limited image quality (score = 2); diagnostic image quality with clinical issues addressed (score = 3); good image quality with no SNR limitation or parallel imaging artifact (score = 4); excellent image quality, or an image quality as good as can be expected (score = 5). The entire exam with different sequences was reviewed and scored on these criteria and on whether the clinical issues were addressed. The mean image quality score was calculated and a confidence interval for the proportion of exams with good image quality was determined.

Results

Phantom studies

Channel decoupling—The phantom images using the 32-channel anterior flexible coil are shown in Fig. 2. Relatively uniform sensitivity of the coil-combined image through the FOV is shown in Fig. 2(b). Excellent channel decoupling of the constructed coil array is demonstrated in Fig. 2(c). The noise correlation matrix (Fig. 2(d)) with the maximum off-diagonal value of 32.7% and the mean correlation value of 6.5% also demonstrates good channel decoupling.

SNR comparison—SNR maps with the 64-channel pediatric coil and the 32-channel standard cardiac coil on the unloaded rectangular phantom, the loaded rectangular phantom, and the loaded pediatric torso phantom are shown in Fig. 3 (a–c) respectively. The corresponding 1D SNR profiles at two selected lines are shown in Fig. 3(d–f). Improved SNR was achieved in all these phantom studies. The SNR gain was depth and load dependent. The averaged SNR gain improved over the standard coil along the center line from the surface of the phantom to a depth of 7 cm was approximately 63%, 37%, and 23% for the unload rectangular phantom, the loaded rectangular phantom, and the loaded pediatric phantom respectively.

Parallel imaging performance—The g-factor maps with different acceleration factors are shown in Fig. 4(b). With the acceleration factor of 3 ($R_y = 3$) in the A/P direction, the averaged g-factor was 1.12 ($R_x = 2$), 1.17 ($R_x = 3$), and 1.38 ($R_x = 4$) for the pediatric coil, and 1.17, 1.36, and 1.91 respectively for the standard coil. With $R_y = 4$, the averaged g-factor was 1.18 ($R_x = 2$), 1.26 ($R_x = 3$), and 1.56 ($R_x = 4$) for the pediatric coil, and 1.36, 1.65, and 2.23 respectively for the standard coil. Better parallel imaging performance (lower g-factor) has been achieved by the dedicated pediatric coil under all different acceleration factors.

In vivo studies

Representative images of a three-year-old male patient are shown in Fig. 5. Using different coil modes, coverage of the full spine and abdomen was achieved by the 64-channel pediatric coil. Supporting Figure S2 shows representative images of a five-year-old patient with a vascular malformation in the hand. Coverage of the entire extremity from shoulder to fingertip was also achieved. The mean score for overall image quality of the 25 exams was 4.6. All exams were at least diagnostic (score no less than 3), and all corresponding clinical

issues were addressed. 72% of the exams achieved excellent image quality (score of 5), and 92% of the exams achieved good image quality (score of 4 and 5). The 80% confidence interval for the rate of obtaining an exam of good or excellent image quality was 85–99%.

Discussion

The lack of dedicated pediatric coils limits the quality and access of MRI for pediatric patients. In this work, we have developed a semi-flexible 64-channel phased array for pediatric body MRI at 3T. The anterior coil is flexible in the transverse plane, and can conform to different patient sizes. Thus, image quality can potentially be improved by the dedicated pediatric coil. Better parallel imaging performance can also be expected compared with adult coils. The large coverage of the 64-channel pediatric coil enables arm, spine, and abdomen/pelvis imaging without coil repositioning. This can improve the scan efficiency and reduce the total scan time for pediatric patients, who often present with a need to image multiple anatomic regions.

Employing $\lambda/4$ Baluns enabled a very deterministic design approach, and produced excellent SNR and channel decoupling. They are easy to implement, which may favor commercialization for widespread use by different institutions and vendors. Note that these concepts are not limited to pediatric body MRI, and can be extended to other coil design and development.

There are several limitations of this work. First, although the anterior coil is flexible and can easily conform to patient sizes, the posterior coil is fixed in the patient table. The curvature of posterior coil is designed to match the shape of small children, and may not be ideal for adults. For the rectangular phantom studies, the curvature of the pediatric coil will cause unnecessary gaps between the phantom and the coil, therefore only the anterior coils were compared for the rectangular phantoms. Careful patient positioning was also required for the pediatric coil since the position of the posterior coil cannot be adjusted. A more flexible posterior coil is currently being designed and constructed.

Another limitation is that the performance of the 64-channel pediatric coil has only been objectively compared to a commercially available 32-channel cardiac coil in the phantom studies. While in vivo imaging with both a conventional coil array and our flexible coil array might permit direct assessment of the clinical impact of the developed coil, performing such a study would involve multiple contrast agent injections and prolonging the duration of anesthesia significantly for the pediatric patients. A volunteer study may be an alternative, but since the coil was designed for pediatric patients, we do not expect significantly better performance of this coil on adult volunteers. The recruitment of cooperative non-sedated pediatric volunteers is arguably impractical. Another alternative is that we could retrospectively identify pediatric patients with similar weights or ages, who were imaged with the same scan protocol using either an adult coil or the flexible coil. An image quality comparison between these two groups may provide indirect assessment of the flexible coil, but the sample size needs to be relatively large to reach a statistically significant conclusion. This will be the subject of future work.

In conclusion, a semi-flexible 64-channel phased array dedicated for pediatric body MRI at 3T has been designed, constructed, and validated. The new coil array conforms to different pediatric patient sizes, provides good image quality and large body coverage, and can facilitate pediatric MRI in various clinical applications.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1.

(a) Coil layout of the 32-channel anterior coil. (b) Circuit schematic for a single coil element. (c) The anterior coil without the mechanical package: the coil elements only overlap in the S/I direction, creating high flexibility in the transverse plane. The cables were routed through the center of the coil arrays to achieve minimum coupling of the coils and cables. Floating cable traps are marked by yellow circles. (d) Demonstration of coil-to-patient matching on a doll for the anterior coils of a standard flexible 32-channel cardiac coil array (left) and the developed 64-channel pediatric coil (right). The anterior coil can automatically conform to the patient.

R1.3: *Floating cable traps marked.*

Figure 2.

$\lambda/4$ Baluns and channel decoupling: (a) self-resonance frequency of the ground loop formed by the attached cables depends on the location of the ground connection. Setting the ground connection at $\lambda/4$ results in highest common mode impedance. That breaks the ground loop and essentially has the same effect as conventional Baluns. A phantom study demonstrates good channel decoupling achieved by the 32-channel anterior coil: (b) images from all coil elements and (c) the corresponding images from all individual coil elements. (d) The noise correlation matrix of the 32-channel anterior coil also demonstrates good channel decoupling.

Figure 3.

SNR comparison between the anterior coils from the 64-channel flexible coil and the standard 32-channel cardiac coil on (a) an unloaded rectangular phantom, (b) a loaded rectangular phantom, and (c) a pediatric shape loaded phantom. The 1D SNR map at two selected locations (highlighted by black lines) are shown in (d), (e) and (f) respectively. The 64-channel flexible coil achieved better SNR in all cases compared to the standard 32-channel coil. The achieved SNR gain over the standard coil is depth and load dependent.

Figure 4.

Comparison of g-factor of the 64-channel flexible coil and the standard 32-channel coil under different acceleration factors. The experiment setups are shown in (a), and the g-factor maps are shown in (b). The 64-channel flexible coil has achieved better parallel imaging performance than the 32-channel standard coil for all acceleration factors. Note that lower torso mode of the 64-channel flexible coil was used.

Figure 5.

A three-year-old male patient with clinical concerns of tethered cord: (a) T1-weighted FSE of cervical and upper thoracic spine and (b) its axial reformat; (c) T2-weighted FSE images of the sacrum; (d) 3D FSE images with an acceleration factor of 6; (e) T2-weighted images and (f) axial reformat; (g) post-contrast T1-weighted images; (h) images after maximum intensity projection show good image quality of the kidney. (a), (b), (c), (e) and (f) were imaged using the posterior mode, and the other images were imaged using the middle torso mode. The array enabled detailed imaging of multiple body parts without repositioning the patient, in this case a full spine, abdomen, and pelvis.