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A Dedicated 36-channel Receive Array for Fetal MRI at 3 T

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Abstract

Due to a lack of fetal imaging coils, the standard commercial abdominal coil is often used for fetal imaging, the performance of which is limited by its insufficient coverage, element number and SNR. In this work, a dedicated 36-channel coil array, of which size can best fit the body sizes of pregnancy gestation from 20 to 37+ weeks, was designed for fetal imaging at 3T. Signal-to-noise ratio (SNR) with full phase encoding and G-factor denoted as noise amplification for parallel imaging were quantitatively evaluated by phantom studies. Compared to a commercial abdominal coil array, the proposed 36-channel fetal array provides not only SNR improvements in full phase encoding (with 10% in the region where the whole fetal body was located, and up to 40% in the edge region where the fetal brain and heart may appear), but also an augmented parallel imaging capability and remarkable SNR improvements at high acceleration factors.

Keywords

Index Terms— fetus imaging; magnetic resonance imaging (MRI); phase array; parallel imaging; radio-frequency (RF) coil

I. INTRODUCTION

MAGNETIC resonance imaging (MRI) has been an important tool for defining fetal anatomy and pathology non-invasively due to its non-ionizing radiation feature, superior soft tissue contrast, and the capability of a large FOV scan to image the entire body of the fetus.

However, because of fetal and/or the mother's involuntary motion, particularly in applications with long scan times, a deterioration in image quality remains a key limiting factor to the use of MRI [1]-[2]. Advances in the development of ultrafast imaging sequences, such as balanced steady state free precession (bSSFP) and half-Fourier acquisition single-shot turbo spin-echo (HASTE) sequences, have been commonly used for fetal MRI [3]-[6]. Both bSSFP and HASTE allow scanning in one imaging plane within a second, and thus the fetal or maternal motion can be frozen during data acquisition. In addition, bSSFP and HASTE provide T2/T1 or T2-weighted images, which makes it

possible to achieve excellent contrast resolution of fetal tissues without the use of fetal sedation or paralysis.

Due to the need for high signal-to-noise ratio (SNR) in the fast imaging acquisition, the utilization of the phased array receive coils with parallel imaging technology [7]-[12] would be essential for the application of ultrafast MRI sequences. In fetal MRI today, the commercially available torso or abdomen coil array combined with a spine array is used in clinical settings [5]. This coil array setup of combined torso coil and spine coil designed for regular adults is not optimized for fetal imaging applications in terms of imaging acceleration capability, efficiency, safety and imaging coverage. Moreover, the image quality in the prenatal period is limited by low SNR. High density (or high channel count) arrays are needed to minimize acquisition time and maximize the SNR.

Many investigations have been performed to obtain a well-designed phased array coil to meet the requirements of specific applications. Previous work indicated that the coils designed with microstrip transmission lines [13]-[21] can be constructed in different sizes and shapes [22]-[23], and these studies show the unique capability of a large coverage and improved flexibility, which are potential for fetal MRI. For SNR improvement and acquisition time reduction in clinical fetal MRI, alternative methods include the use of increased magnetic field strength [24]-[26], and an increased element number of the phased array receive coils with small coil elements [27]-[28]. Based on these studies, some dedicated coils have demonstrated for highly accelerated fetal imaging by using a receive coil with a large channel number and small sizes of coil elements [29][33]. Moreover, flexible coils have the potential to further improve the image quality [34], such as the flexible highly configurable 16-channel array coil [29]. Compared to a regular rigid coil design, flexible coils can tightly conform to the anatomy of imaging subjects, leading to significantly reduced distance between coil and subjects and also improving filling factors, which can improve detection sensitivity and excitation efficacy [35]. However, the 16-channel coil array was designed for a 1.5 T MRI system with fewer channels. And the semi-adjustable 64 channel array coil was only designed for fetal imaging at 22 weeks of pregnancy, though it has a highly accelerated performance [33]. Moreover, the technical issues related to the coil design with higher channel count, including elements decoupling, noise correlation, loading, and common-mode rejection, become more challenging. Therefore, to conform to different pregnant women sizes and to achieve fast scanning with high accelerated capability in the clinical diagnosis, a new coil array design need to be taken into consideration.

In this study, we designed and built a 36-channel receive coil array dedicated for fetal MRI at 3T with wide coverage and flexibility to conform to the varied anatomy of pregnant patients. The performance of the coil array was characterized with improvements in SNR, parallel imaging capability, and image quality compared to that of a commercially available 18 channel abdomen coil array.

II. METHODS

A. Coil design

RF coil arrays designed for abdomen and torso MR imaging, are often unable to completely conform to the varied anatomy of pregnant patients, leaving some receive elements distant to the abdomen, consequently reducing the detection sensitivity.

Before the coil was designed, the body sizes of different pregnancy gestations were measured when the women were lying supine, which is the position for medical diagnosis. The measurement sketch of the body sizes is shown in Figure 1, and some of the results are depicted in Table I.

Most 3 T systems allow 48 independent receiver channels that can be used simultaneously in one single scan and in one single FOV, each generating an independent partial image. For fetal MRI, a 12-channel coil in the posterior position (generally four channel in one row, making a total of three rows) from the 32-channel spine array is used for imaging, combined with an 18-channel abdomen coil, totaling 30 channels. This limits the acceleration factors that can be achieved. Therefore, to obtain high acceleration factors, the channel number of the dedicated coil can be chosen as 36. Combined with the existing 12-channel spine coil, 48 channels can be achieved.

Hence, to cover pregnancy gestation from 20 to 37+ weeks, a dedicated 36-channel coil array for fetal imaging was designed with sizes 386 mm $\times$ 761.5 mm. The coil layout and the prototype coil array are shown in Figure 2. The 36 coil elements were printed on polyimide flexible circuit material, using 6 mm-wide, 1 oz/ft² Cu traces in a rectangular shape, and were arranged in four rows in the superior/inferior (S/I) direction, with nine coils on each row in the left/right (L/R) direction.

The sizes of channels 2, 4, 33, and 36 were each 98 mm $\times$ 56.5 mm and the size of each of the others was 110 mm $\times$ 103 mm. Here, the decoupling methods of overlapping adjacent coil elements and employing low input-impedance preamplifiers were applied in the design. Therefore, the distance between the elements needs to be carefully designed. Each element was tuned to 123.2 MHz and impedance-matched to minimize the noise of the preamp. Additionally, two baluns at 123.2 MHz were placed in the two large plugs, which were spiral and self-shielded, to minimize noise. To maintain flexibility and gain durability and safety, the coil array was covered by the polyethylene material.

Figure 3 shows the circuit schematics of the coils, in which channels 2, 4, 33, and 36 in a small size were divided into four segments and the other channels were divided into six segments by the capacitive breaks. The matching and tuning circuits, respectively, were composed of a parallel circuit with a constant capacitors (100B, ATC) and an adjustable capacitors (JZ150, Voltronics). In the loop, the active detune circuit was comprised of a capacitor C1, an inductance L1 (32 – 39 nH), and a PIN diode (DH80055–40N, TEMEX). Additionally, a passive-detune circuit including an inductance L2 (32 – 39 nH) and an antiparallel diode pair (UMX9989AP, Microsemi), were added parallel to the capacitor C2.

To minimize the noise of the preamp, an RF choke and a capacitor C3 were placed in the coils, as shown in Figure 3.

B. Bench test

Bench testing was conducted with the coil element tuning and matching, active detuning, and decoupling with the nearest-neighbor element by using a custom-made test power supply and a vector network analyzer (Agilent, E5061B).

The power supply can not only provide 10 V for the preamplifiers, but also detune the transmit coil and receive coil with a switch to bias PIN diodes with a 100-mA current during the transmit mode, and reverse biasing of the diodes with -30 V during image acquisition.

The tuning and matching of the individual elements was set by monitoring the S11 parameter. To perform the decoupling, the S21 parameter was measured with cables directly connected to the preamplifier sockets of the two adjacent or nonadjacent coil elements by the vector network analyzer, when all other unused coil elements were actively detuned with the PIN diode under forward bias and had the preamps attached to supply preamplifier decoupling. The coil quality factor Q was calculated by the equation:

$$Q = f_0/\Delta f \quad (1)$$

where f_0 is the center frequency and f is -3 dB bandwidth (BW). In the vector network analyzer, the quality factor Q of an individual element with no cable attached can be measured directly using an S12 measurement between two shielded inductive probes loosely coupled to the coil element.

The loaded quality factor was tested with a body-sized phantom (35 cm wide, 26 cm thick and 45 cm length) capable of electrical properties mimicking that of human tissue. The phantom was filled with 1.24 g/L NiSO₄·6H₂O and 2.62 g/L NaCl (approximate conductivity 0.65 S/m and relative permittivity 83 at 123.2 MHz, measured by the vector network analyzer) to simulate the whole body's average dielectric properties.

C. Phantom Studies

Phantom experiments of the 36-channel coil were first implemented on a 3-T clinical MRI Siemens scanner (Magnetom Skyra, Siemens Healthcare, Erlangen, Germany) in comparison with a commercially available 18-channel abdomen coil array (Body 18, with dimensions 385 mm $\times$ 590 mm), which is generally used for fetal imaging by the doctors. In the comparison experiments, the 36-channel coil and the body 18 in the anterior position were both combined with a 12-channel coil in the posterior position from the 32-channel spine array in the patient table (Spine 32). Phantom signal images were acquired with a 2D GRE sequence (TR/TE = 300 ms/10 ms, flip angle = 60° , FOV = 400 mm, slice thickness = 5 mm, and matrix size = 256×256), and the noise images were obtained from the same GRE sequence without RF excitation. The results can be applied to compute covariance weighted-SNR [36], SENSE g-factor maps, and SENSE SNR maps [37], and the calculation formulas are as follows:

$$SNR^{cov - rSoS} = \sqrt{S^H \Psi^{-1} S} \quad (2)$$

$$g_\rho = \sqrt{\left[(S^H \Psi^{-1} S)^{-1} \right]_{\rho, \rho} \cdot (S^H \Psi^{-1} S)_{\rho, \rho}} \quad (3)$$

$$SNR_{SENSE} = \frac{SNR_{full}}{g_\rho \cdot \sqrt{R}} \quad (4)$$

where S denotes the signal image (H indicates transposed complex conjugate), Ψ is the noise matrix, R represents the acceleration factor, and SNR_{full} is the SNR with full phase encoding.

To assess the fetal coil array performance, the mean values of the combined SNR maps and the inverse g-factor maps were calculated in several ROIs where the fetal brain and heart may appear for their motility.

D. In vivo studies

Generally, the pregnant women would lie in the supine position during MRI scanning and the fetal anatomy position seen in the images are consistent with the doctor's diagnostic habits. If the mother does not tolerate the supine position, the examination can be performed in the decubitus position [2].

A 2D HASTE sequence was applied to obtain fetal brain images with the following parameters: TR/TE = 1400 ms/92 ms, FOV = 380 mm x 308 mm, slice thickness = 3.5 mm, slices = 20, flip angle = 160°, matrix size = 320 × 234, bandwidth = 710 Hz/Px, accelerate factor = 2, and phase partial Fourier = 5/8. For the fetal images reconstructions, the standard image reconstruction system was used on the scanner.

In the in-vivo study, both the 36-channel coil and the body 18 were also combined with a 12-channel coil from the spine 32. Notably, a high dielectric mat was placed beneath the pregnant woman during all the fetal imaging to reduce energy deposition [38].

This study was approved by institutional review boards of Shenzhen Institutes of Advanced Technology of Chinese Academy of Sciences. In addition, informed consent was obtained from all participants.

III. RESULTS

A. Bench test results

The average S11 parameter was approximately −18 dB when the coil was loading with the phantom and was −12.5 dB in the worst-case situation. The decoupling between two coil

elements measured by the S21 parameter had an average value of -15 dB, of which the worst-case situation is -12.2 dB for the nonadjacent elements channel 10 and channel 15 with the decoupling method of employing low input-impedance preamplifiers. These indicated that the dedicated coil met the imaging requirements.

In addition, the isolation between the tuned and detuned states of the active detuning was nearly 40 dB, which indicated that the detuning was satisfactory.

Due to different sizes of the 36-channel coil elements, as shown in Figure 2 a, the Q-factor of five representative neighboring coil units (channels 28, 31, 33, 34, and 36) was tested with loading and unloading of the body-size phantom, in which the size of channels 33 and 36 was nearly half the size of the other channels. The unloaded-to-loaded quality factor ($Q_{\text{unload}}/Q_{\text{load}}$) was 95.5/11.9 (channel 28), 115.0/15.5 (channel 31), 105.8/32.1 (channel 33), 102.4/13.4 (channel 34), and 103.8/21.8 (channel 36).

B. Phantom studies

To validate the coil decoupling, the noise correlation matrix was calculated and is shown in Figure 4 (the mean values for all except the diagonal elements are also displayed). The noise correlation of the 36-channel coil revealed a mean of 7.3%. For the 18-channel coil, the mean was 5.7%. Although the mean of the noise correlation matrix of the 36-channel coil was slightly larger than that of the 18-channel coil, it demonstrated good channel decoupling.

The SNR distributions with full-phased encoding from phantom images in the transverse, sagittal and coronal planes are shown in Figure 5. In the first and second row maps, the mean SNR values in a larger ellipse region of interest (ROI), where the whole fetal body was located, and in four small circle ROIs where the fetal brain and heart may appear, were illustrated in the corresponding positions. Compared to the 18-channel abdominal coil, the 36-channel fetal coil achieved an average SNR improvement of 10% in the large ROI where the whole fetal body was located, at least 6% in the center ROI and up to 40% in the edge region (in the sagittal and coronal planes). Additionally, the 1D SNR variations in the regions indicated by the white lines were shown in the third row of the figure. From the profiles, it can be deduced that the SNR in the edge region of the 36-channel coil was much higher than that of the commercial abdominal coil, and at the center region, the 36-channel coil had nearly the same SNR as the 18-channel coil.

Measurements of the transmitted RF field with (489.63 V) and without (498.21 V) the constructed detuned array coils inside the scanner showed up to a 2% difference in the transmit power. Moreover, the difference in the transmit power between with the 36-channel coil (489.63 V) and with the 18-channel coil (483.32 V) was 1.3%. This indicates sufficient coil element detuning and suppression of common mode currents on the cables.

Figure 6 depicts the inverse g-factor maps in the transverse plane with acceleration factors ranging from 2 to 8 in the L/R direction. Known from the mean 1/g-factor values and the peak g-factor values illustrated in the maps, the proposed 36-channel coil showed a better parallel imaging capability than that of the 18-channel coil, particularly at high acceleration

factors. For fetal MRI, the most commonly used accelerator is 2, and it would be 4 in fetal heart imaging. Here, we just evaluate the parallel imaging performance of the coils by the simulation with phantom results.

The SNR improvements of the proposed 36-channel coil compared to the commercial abdominal coil in the transverse plane with different acceleration factors in the L/R direction are shown in Figure 7, which demonstrates that the 36-channel coil can achieve a higher peripheral SNR and show substantially better SNR performance in all areas with higher accelerated acquisitions compared to the 18-channel array coil.

C. In vivo studies

The fetal images obtained from a 24-week pregnant woman at the 3 T MRI system are shown in Figure 8. Fetal images are usually acquired in the transverse, sagittal, and coronal planes, orthogonal to the fetal anatomy. For the transverse and coronal planes, the acceleration is in the AP direction, and it is the RL direction for the sagittal plane, passing through a suitable fetal head position. Note that the fetal brain, spinal cord, liver, and extremities can be seen clearly, as depicted in the sagittal images. In the figure, a = brain, b = liver, c = spinal cord, d = extremities, and e = placenta.

Figure 9 shows an example with the magnified brain fetal images in the transverse plane, in which the numbers denote the mean signal values in the ROIs at the similar noise level for the both coils, dividing the absolute signal values by the mean noise of the three regions, as shown in the transverse plane in Figure 8. Known from Figure 9, the proposed 36-channel fetal coil had higher signals than that acquired by using the commercial abdominal coil, particularly in the fetal brain area. As a result, a conclusion can be drawn that the fetal images acquired by the 36-channel coil were of better quality than those of the 18-channel coil.

IV. DISCUSSION AND CONCLUSION

In this work, we have developed a 36-channel array coil for fetal imaging at 3T. The coil is flexible, and it may conform to different patient sizes, as shown in Table I. Compared with an 18-channel abdominal coil, the dedicated 36-channel fetal coil has a larger imaging FOV, and much higher SNR over the surface area, as demonstrated by the phantom images in the transverse plane. For a larger number of coil elements, and a higher element density, better imaging acceleration capacity also can be achieved. Particularly, higher SNR improvement in the edge region can be produced with a high acceleration factor, known from Figure 7. As a result, better fetal image quality can be achieved by using the 36-channel fetal coil array.

As the proposed coil array provides good image quality and large body coverage, it can facilitate fetal MRI in various clinical applications. Generally, the more coil elements with a higher element density that can be used, the better the image quality and acceleration capacity that can be achieved in parallel imaging. As the dedicated 36-channel coil array has many more coil elements than the 18-channel abdominal coil, it should be useful for some special applications, such as real-time heart imaging, in which parallel imaging with a high

acceleration factor is necessary. Nevertheless, further studies need to be done for the fetal coil array in improvement of the imaging acceleration capacity and SNR.

In conclusion, a dedicated 36-channel array coil for fetal imaging was designed, constructed, and evaluated by imaging experiments in phantoms and pregnant women. In comparison with a commonly-used 18-channel abdominal coil for fetal imaging, the dedicated 36-channel fetal coil achieves a better performance in MR SNR, imaging acceleration capacity, and in the quality of the fetal images, demonstrating its potential for fetal clinical diagnosis.

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Fig. 1.
Measurement sketch of the body sizes.

a. Electrical layout of the 36-channel fetal coil; b and c. Photographs of the 36-channel coil.

Fig. 3. Circuit schematics of the coil elements: a. the circuit of channels 2, 4, 33, and 36 in a small size with four segments by the capacitive breaks; b. the circuit of the other channels with six segments by the capacitive breaks.

Fig. 4. Noise correlation matrix of the proposed 36-channel coil and the commercial abdominal coil both combined with a 12-channel spine coil. The mean off-diagonal of the matrix is shown in the corresponding maps.

Fig. 5. SNR distributions of the phantom results in the transverse, sagittal and coronal planes. The mean SNR values of the ROIs depicted as the circles are shown in the corresponding positions. The curves in the third row illustrate the SNR variations in the regions indicated by the white lines. Directions: anterior/posterior (A/P), left/right (L/R), superior/inferior (S/I).

Fig. 6. 1/g-factor maps of the phantom results in the transverse plane with acceleration in the L/R direction. The mean 1/g-factor values in the ellipse ROIs are displayed in the corresponding positions. Peak g-factor values for each case are shown below each map.

Fig. 7. SNR improvements of the proposed 36-channel coil compared to the commercial abdominal coil in the transverse plane with different acceleration factors in the L/R direction of the phantom results.

Fig. 8.

Fetal images of a 24 weeks pregnant woman with a supine position in the transverse, sagittal and coronal planes acquired by using the proposed 36-channel coil and the commercial abdominal coil both combined with a 12-channel spine coil. All the images are displayed at the same window level. a = brain, b liver, c = spinal cord, d = extremities, and e = placenta. The numbers in the circles represent the mean noise in the corresponding regions.

Fig. 9.

Magnified brain fetal images in the transverse plane acquired by using the 36-channel coil and the commercial abdominal coil both combined with a 12-channel spine coil, shown at the same window level. The numbers denote the mean signal values in the ROIs at the similar noise level, dividing the absolute signal values by the mean noise of the three regions, as shown in the transverse plane in Figure 8.

TABLE I

The Body Sizes of Different Pregnancy Gestations (Lying Supine)

Gestational weeks	18	23	24	24	28	36
1. width of abdomen (cm)	48	52	56	65	70	74
2. width of back (cm)	30	30	26	36	29	25
3. up size from navel (cm)	10	10	14	10	16	21
4. down size from navel (cm)	15	15	15	15	19	16
5. width of the body side (cm)	19	21	27	31	31	35