Investigation of multichannel phased array configurations for fetal MR imaging at 1.5T

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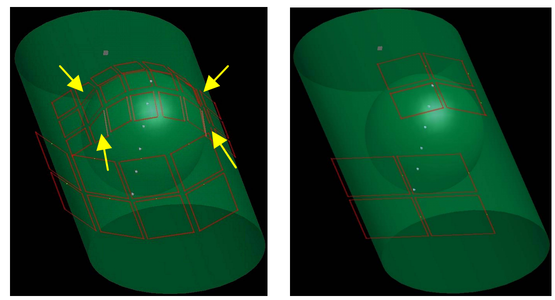
Introduction Fetal MRI on 1.5T clinical scanner has increasingly been performed to evaluate the brain abnormalities and the potential of neurodevelopment disabilities since its first introduction in early 1980s [1]. Due to the susceptibility to the fetal motion, mutltiecho ultrafast MRI techniques such as single-shot fast spin-echo (ssFSE) and half-Fourier acquired ssFSE are primarily used to reduce the motion artifacts with signal-to-noise ratio (SNR) scarification, which demands SNR improvement by hardware. In addition, since there is few dedicated fetal phased array [2], commercial torso or cardiac phased arrays are routinely used instead, which provide unoptimized SNR for fetal imaging due to the limited coil elements, coverage and filling factor. Current research has shown the dramatic potential to improve SNR of region near the coil as well as deeper region in a spherical model [3, 4] by increasing the number of coil elements. This work proposes a flexible 32-channel fetal phased array design to increase SNR of the whole uterus region since the head of fetus can be anywhere with in the uterus. The array performance is simulated by using Finite-Difference Time Domain (FDTD) method and compared with a commercial 8-channel torso array at 1.5T.

Methods As shown in Fig. 1(a), the dedicated 32 channel fetal array consisted of 4×2 square surface coils with 110 mm width and 160 mm length on the bottom and 8×3 coils with 60 mm width and 70 mm length at the top except the four trapezoidal coil indicated by yellow arrows. By increasing the number of coil element and the relative small size of each element, the array is more flexible for different patients as well as the same patient in different trimesters. Compared with the 8-channel commercial torso array, which consisted of 4 square surface coils with 160 mm width and 160 mm length on the bottom and the other four with 110 mm width and 110 mm length at the top as shown in Fig 1(b), the coverage and filling factor were improved along with the flexibility increased.

The simulations of the two arrays were carried out using commercial FDTD software XFDTD 6.5 (REMCOM, Inc., State College, PA) to compare array performance. The conductors (red region) were copper tapes ($\sigma = 5.8\times10^7$ S/m, $\mu_r = 1$ and 3 mm in width). The phantom (green region, $\sigma = 0.7$ S/m and $\mu_r = 72$) was ellipse cylinder with 800 mm length, 205 mm long axel and 120 mm short axel, combined with a sphere with 140 mm radius. In order to achieve better coverage and filling factor, the coil elements at the top of the torso array were rotated 15º along the anterior-posterior direction. All the elements of the proposed fetal array were placed close to the phantom. A three-dimension FDTD simulation was performed at 64 MHz, corresponding to the proton Larmar frequency at 1.5T. Each element of the two arrays was excited by sinusoidal current source with RMS value of 1 A and the same phase. Outer boundaries were absorbing perfectly matched layer (PML) with 7 layers. The meshing cells of the two models were 3mm×3mm×5mm.



(a) 32 channel fetal array (b) 8 channel torso array
Fig. 1 Configurations of coils and phantoms

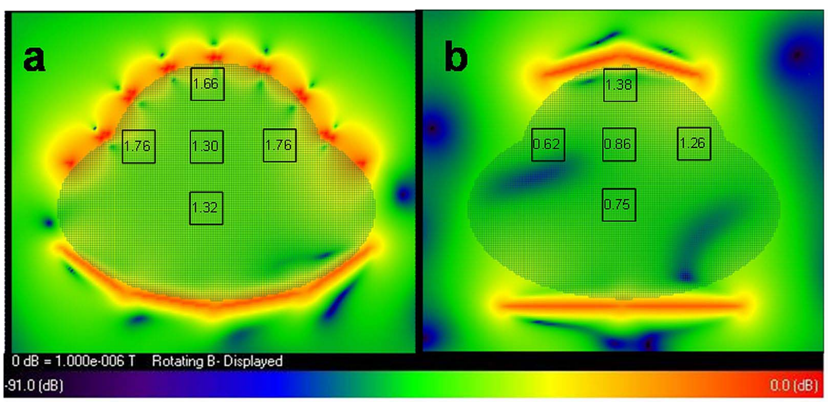


Fig. 2 B_1 map of (a) 32 channel fetal array and (b) 8 channel torso array in the central transversal plane of the phantom calculated by XFDTD. The numbers in the boxes indicated the mean B_1 (10^{-8}T) in the 3cm × 3 cm region.

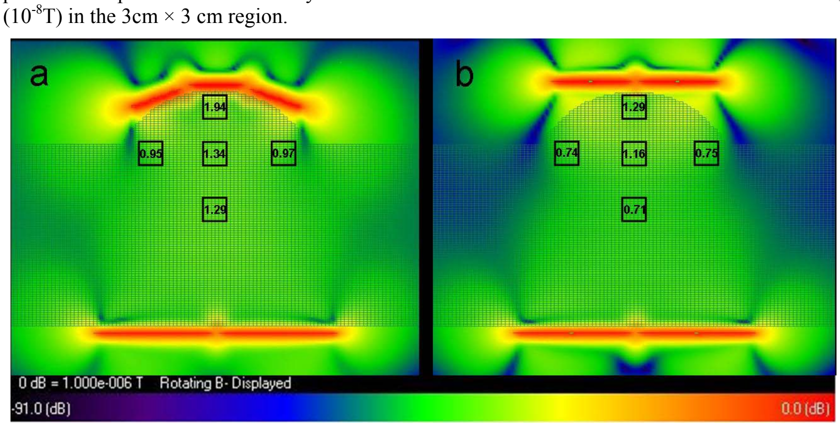


Fig. 3 B_1 map of (a) 32 channel fetal array and (b) 8 channel torso array in the central sagittal plane of the phantom calculated by XFDTD. The numbers in the boxes indicated the mean B_1 (10^{-8}T) in the 3cm × 3 cm region.

Results The B_1 field distribution in the transversal and sagittal planes of the two arrays, which was scaled to 2×10^{-7} W input power of each element, was shown in Fig. 2 and Fig. 3. The mean B_1 in 3cm × 3cm region at different location in the whole uterus was shown in the black boxes. As shown in Fig. 2, B_1 was increased 20% in the surface region at the center of transversal plane, whilst that on left and right sides increased 40% to 180% due to the better coverage of the 32 channel fetal array. As shown in Fig. 3, B_1 increased 50% in the center of surface region as well as that on the anterior and posterior sides was increased 28% due to better filling factor of the fetal array. B_1 in the center of uterus and in deeper region such as the center of the patient was increased 87% and 79% respectively because of increasing the number of element. Besides the improvement of B_1 field strength, the B_1 homogeneity also increased substantially which is important for fetal MRI due to the possibility of fetus head location in the whole uterus.

Discussion There are some visible dark spots of the B_1 distribution of the fetal array at the surface of phantom as shown in Fig. 2 and Fig. 3 which may be due to the magnetic field cancellation of adjacent elements. This limitation can be compensated by simple post-processing algorithms of image reconstruction, such as fine adjusting the amplitude and phase of the signal of each element. The simulation results indicate the proposed 32 channel fetal array improves SNR and B_1 homogeneity by increasing the number of coil element. Future work will focus on parallel performance evaluation of the fetal array.

References

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