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Simulation verification of SNR and parallel imaging improvements by ICE-decoupled loop array in MRI

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Abstract

Transmit/receive L/C loop arrays with the induced current elimination (ICE) or magnetic wall decoupling method has shown high signal-to-noise (SNR) and excellent parallel imaging ability for MR imaging at ultrahigh fields, e.g., 7 T. In this study, we aim to numerically analyze the performance of an eight-channel ICE-decoupled loop array at 7 T. Three dimensional (3-D) electromagnetic (EM) and radiofrequency (RF) circuit co-simulation approach was employed. The values of all capacitors were obtained by optimizing the S-parameters of all coil elements. The EM simulation accurately modeled the coil structure, the phantom and the excitation. All coil elements were well matched to 50 ohm and the isolation between any two coil elements was better –15 dB. The simulated S parameters were exactly similar with the experimental results, indicating the simulation results were reliable. Compared with the conventional capacitively decoupled array, the ICE-decoupled array had higher sensitivity at the peripheral areas of the image subjects due to the shielding effect of the decoupling loops. The increased receive sensitivity resulted in an improvement of signal intensity and SNR for the ICE-decoupled array.

Keywords

Magnetic resonance image (MRI); magnetic wall; induced current elimination (ICE); decoupling; RF coil; phased array; SNR; parallel imaging

Introduction

Ultrahigh field MRI (7 T and higher) promises high signal-to-noise (SNR), high spatial resolution and high temporal resolution in MR imaging and spectroscopy [1–4]. However,

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several challenges for RF engineering exist at ultrahigh fields, e.g. radiofrequency (RF) field non-uniformity and increased special absorption rate (SAR). Commercial 7T scanners have not been equipped with the body coil, which is a standard configuration in 3T and lower magnetic field. Therefore the RF coil used for ultrahigh field MRI should exhibit the ability of both transmission and reception. Compared with transmit-only coil and receive-only coil configuration, transmit/receive (transceive) coils do not need detuning circuits and are easier to fabricate in practice [5–9]. In addition, transceiver coils with array configuration exhibit the capability of RF shimming, parallel transmission (pTx) and parallel imaging.

In the transceive array design, one of the main challenges is to attain sufficient electromagnetic isolation among array elements. Several decoupling methods have been employed for RF arrays, e.g. element overlapping [10], transformers [8] and L/C networks [11,12]. A new decoupling method, namely the induced current elimination (ICE) or magnetic wall decoupling method, has been recently proposed [13] and successfully applied in microstrip line [13,14], L/C loop [15,16], monopole [17] and dipole arrays [18]. The ICE decoupling used external resonators to compensate the induced current caused by electromagnetic (EM) coupling. Compared with traditional decoupling method, ICE decoupling is much more robust and exhibit superior decoupling performance in both loaded and unloaded cases [15].

From experimental results [15], it is found that ICE-decoupled loop array has higher SNR and better parallel imaging performance over capacitively decoupled array. In this study, we aim to verify the improvement of SNR and parallel performance of the ICE-decoupled loop array through rigorous numerical simulations. Two eight-channel loop arrays using capacitive decoupling and ICE decoupling were numerically investigated. Three dimensional (3-D) EM and RF circuit co-simulation approach [19–21] was employed because of its fast speed and accurate results. Decoupling performance, B_1 map, SNR and parallel imaging performance of the two arrays were quantitatively evaluated and compared.

Methods and Materials

In this study, numerical studies were conducted by a full-wave EM simulation software HFSS (ANSYS, Canonsburg, PA, USA), which is based on the finite element method (FEM). Compared with other software based on finite-difference-time-domain (FDTD) method, HFSS is more flexible for modeling arbitrary geometries because it can be applied to a non-structured grid. Therefore HFSS can generate accurate results and meanwhile do not require long computational times. In addition, HFSS has a robust approach for mesh refinement and coverage criteria, which further improves the simulation accuracy and reliability [22]. The limitation of HFSS relative to the FDTD-based software is that it requires much more memory. In this study, a powerful computer with 128G RAM was used for simulation, which can handle the project with a mesh of more than 3 million tetrahedral, adequate for an 8-channel head coil.

The eight-channel ICE-decoupled array was mounted on a cylindrical former with an outer diameter of 25 cm, as shown in Figure 1A. Each rectangular loop element using copper conductor had a dimension of 17 cm $\times$ 6.8 cm, as shown in orange color in Figure 1A. Six

capacitors (one matching capacitor, one tuning capacitor and four distributed capacitors) were equally distributed along the patch of each loop element. Smaller rectangular loops with six distributed capacitors (referred to as the decoupling capacitors) were placed between adjacent loop elements for decoupling, referred to as the decoupling loops, as shown in yellow color in Figure 1A. The dimension of the decoupling loop was $17 \text{ cm} \times 2.8 \text{ cm}$. The width of all copper conductors was 5 mm. A cylindrical water phantom with a length of 37 cm and an outer diameter of 16 cm was placed in the center of the coil array. The electromagnetic parameters of the water phantom were set as follows: conductivity $\sigma = 0.59 \text{ S/m}$; relative permittivity $\epsilon_r = 78$.

As comparisons, an 8-channel capacitively decoupled loop array and a single loop were also modeled in simulation, as shown in Figure 1B and Figure 1C. The dimensions of the capacitively decoupled array and the single loop were exactly the same as that of the ICE-decoupled array. All loop elements were matched to 50 ohm and tuned to 297.2 MHz, which is the Larmor frequency of the 7T MR scanner.

Values of all capacitors were obtained by EM and RF circuit co-simulation method [19,21]. During circuit optimization, all capacitors are variables and their values were determined by minimizing the S-parameters (S_{xx} and S_{xy}). The error functions of the optimization were:

$$EF = \sum_i w_{xx,i} \times |S_{xx,i} - S_{xx,Target}|^2 + \sum_j w_{xy,j} \times |S_{xy,j} - S_{xy,Target}|^2,$$

where i and j are the individual element and the decoupled pair of the 8-channel array, respectively; $w_{xx,i}$ is the weighting factor for the coefficient (S_{xx}) of the individual array element “ i ”; $w_{xy,j}$ is the weighting factor for the transmission coefficient (S_{xy}) of the decoupled pair “ j ”; $S_{xx,Target}$ and $S_{xy,Target}$ are the target S parameters of S_{xx} and S_{xy} , respectively. The weighting factors and target S parameters were set as follows: $w_{xx,i} = w_{xy,j} = 1$; $S_{xx,Target} = -45 \text{ dB}$ and $S_{xy,Target} = -30 \text{ dB}$. Simulations were run on a 12-core workstation with a 128 G RAM. All of the simulations cost about a week.

Reception field (B_1^-) of each coil elements was extracted from simulation by Eq. 1 [23]. The 8×8 noise matrixes (Ψ) of the two arrays are calculated by Eq. 2 [24,25], where E_{km} is the local electric field of voxel k from channel m , σ_k is the local conductivity of voxel k , x , y and z are the voxel size in x , y , and z directions. In this study, the voxel size for noise matrix calculation in x , y , z directions is 2 mm, 2 mm and 4 mm, respectively. The SNR here is calculated and normalized by Eq. 3 [26]. To demonstrate the parallel imaging performance of the two arrays, g-factor maps with acceleration of $2\times$, $3\times$ and $4\times$ are calculated and compared. The g-factor results are calculated following standard definition in [27].

$$B_1^- = \frac{(B_x - iB_y)^*}{2} \quad [1]$$

$$\Psi_{mn} = \Delta x \Delta y \Delta z \times \sum_k \sigma_k (E_{km} E_{kn}^*) \quad (m, n=1, 2, \dots, 8) \quad [2]$$

$$SNR = \sqrt{B_1^{-H} * \Psi^{-1} * B_1^{-}} \quad [3]$$

Results

Simulated S-parameter vs. frequency, i.e., reflection coefficients (S_{xx} , $x=1, 2, \dots, 8$) and transmission coefficients (S_{xy} , $x, y=1, 2, \dots, 8$, $x \neq y$), of the 8-channel ICE-decoupled array and capacitively decoupled array were plotted in Figure 2A and Figure 2B, respectively. The simulated reflection coefficient S_{xx} of each loop element was better than -30 dB, indicating all loop elements were well matched to 50 ohm. Average transmission coefficients S_{xy} between adjacent, next adjacent, next next adjacent and opposite loop elements of ICE-decoupled array were -29.1 dB, -17.5 dB, -27.3 dB and -19.1 dB, respectively; whereas those of the capacitively decoupled array were -21.3 dB, -14.3 dB, -20.3 dB and -17.1 dB, respectively. The S-parameter results from simulation are in an excellent agreement with the measured result in our previous work [28], which also indicated the simulation results were accurate and reliable. It is worth noting from Figure 2 that the loop elements of ICE-decoupled array have a narrower -3 dB bandwidth (f_{3dB}) over the capacitively decoupled array. In other words, loop elements of the ICE-decoupled array have higher quality factor (Q-factor, calculated by $2 \times f_{Larmor} / f_{3dB}$), which has also been found in our experimental results [15].

Figures 3A and 3B show normalized noise covariance matrixes of the ICE-decoupled and capacitively decoupled array, respectively. Similar to S-parameter results (Figure 2), the ICE-decoupled array has lower noise covariances compared with the capacitively decoupled array. Mean noise covariances between adjacent, next adjacent and opposite loop elements of ICE-decoupled array were 0.088, 0.133, 0.036 and 0.049, respectively; whereas those of the capacitively decoupled array were 0.071, 0.161, 0.083 and 0.076.

We also evaluated the B_I field and magnetic field (H field) of a single loop, a single element of the 8-channel capacitively array and a single element of the 8-channel ICE-decoupled array, as shown Figure 4. The B_I field and H field were normalized to 1 W power input. From Figure 4A and Figure 4B, we can find that the B_I field of capacitively decoupled array was a little weaker than that of the single loop. This is probably due to the limited decoupling of the capacitively decoupled array, which led to part of the power was transferred to other loop elements, as shown in Figure 4D and Figure 4E. For the ICE-decoupled array, however, the B_I field was even stronger than that of a single element at the peripheral areas of the phantom, as shown in Figure 4A and Figure 4C. This can be attributed to the shielding effect of decoupling elements, which focuses the magnetic field of coil elements and generates locally stronger H field, as shown in Figure 4F. The shielding effect might be caused by the induced current of decoupling loops, as shown in Figure 5.

Figure 6A and Figure 6B show the combined normalized SNR maps of the capacitively and ICE-decoupled array. The ICE-decoupled array has a higher SNR at the peripheral areas and a little lower SNR at the central area. Figure 7 shows the g-factor maps with reduced factor (R) of 2, 3 and 4. Based on the g-factor results, the ICE-decoupled array exhibits better parallel imaging performance over capacitively decoupled array.

Discussions and Conclusion

The performance evaluation and comparison between ICE-decoupled array and capacitively decoupled array has been investigated in our previous work [15]. To validate the experimental results and gain better understanding of the theoretical basis of the coil performance improvement, these two arrays were numerically analyzed and compared. Instead of using the body/human models, a phantom was preferred due to its homogeneity, which makes coil performance easier to compare without interface of tissue heterogeneities and shorter computing times. To achieve reliable results, 3D EM and RF circuit co-simulation approach was applied [19].

The S-parameter plots vs. frequency were in a good agreement with the experimental results, which also indicates the simulation results are reliable and accurate. Similar to the simulation results, it is found that the ICE-decoupled array exhibits better decoupling performance, higher overall SNR and better parallel imaging performance over capacitively decoupled array.

For ICE decoupling approach, the induced current of the decoupling element was used to compensate for the induced current caused by EM coupling between coil elements, as show in Figure 5. It is found that the direction of the induced current of decoupling element is reverse to that of the coil element. This feature makes the decoupling element act like a “magnetic wall” and slightly improves the SNR at the peripheral areas of imaging subject (Figure 4). In other words, the decoupling element could be considered as a shield and has “shielding effect”. Similar to the traditional shield using perfect electric conductors (PEC), the ICE decoupling approach could also improve the Q-factors in both loaded and unloaded cases. However, the ICE method has much better decoupling performance compared with the traditional shield technique. It should be noted that the ICE-decoupled array has relative lower SNR at the central area (Figure 6). This can be understood that the magnetic field of coil elements is somehow blocked by the “shielding effect” and thus the B_1 penetration is slightly decreased.

The shielding effect of decoupling elements slightly improves the B_1 field at the peripheral areas, but has little influence on the overall B_1 profile. This is reasonable given that the decoupling element is a relatively small L/C loop and thus has very shallow RF penetration.

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References

1. Yacoub E, Shmuel A, Pfeuffer J, Van De Moortele PF, Adriany G, Andersen P, Vaughan JT, Merkle H, Ugurbil K, Hu X. *Magn Reson Med*. 2001; 45:588–594. [PubMed: 11283986]
2. Vaughan JT, Garwood M, Collins CM, Liu W, DelaBarre L, Adriany G, Andersen P, Merkle H, Goebel R, Smith MB, Ugurbil K. *Magn Reson Med*. 2001; 24:24–30. [PubMed: 11443707]
3. Collins CM, Smith MB. *Magn Reson Med*. 2001; 45:684–691. [PubMed: 11283997]
4. Cao Z, Park J, Cho ZH, Collins CM. *J Magn Reson Imaging*. 2014
5. Zhang X, Ugurbil K, Chen W. *Magn Reson Med*. 2001; 46:443–450. [PubMed: 11550234]
6. Zhang X, Ugurbil K, Chen W. *J Magn Reson*. 2003; 161:242–251. [PubMed: 12713976]
7. Adriany G, Van de Moortele PF, Wiesinger F, Moeller S, Strupp JP, Andersen P, Snyder C, Zhang X, Chen W, Pruessmann KP, Boesiger P, Vaughan T, Ugurbil K. *Magn Reson Med*. 2005; 53:434–445. [PubMed: 15678527]
8. Avdievich NI. *Appl Magn Reson*. 2011; 41:483–506. [PubMed: 23516332]
9. Alagappan V, Nistler J, Adalsteinsson E, Setsompop K, Fontius U, Zelinski A, Vester M, Wiggins GC, Hebrank F, Renz W, Schmitt F, Wald LL. *Magn Reson Med*. 2007; 57:1148–1158. [PubMed: 17534905]
10. Roemer PB, Edelstein WA, Hayes CE, Souza SP, Mueller OM. *Magn Reson Med*. 1990; 16:192–225. [PubMed: 2266841]
11. Wu B, Zhang X, Qu P, Shen GX. *J Magn Reson*. 2006; 182:126–132. [PubMed: 16829145]
12. Wu B, Zhang X, Qu P, Shen GX. *Magn Reson Imaging*. 2007; 25:418–424. [PubMed: 17371734]
13. Li Y, Xie Z, Pang Y, Vigneron D, Zhang X. *Med Phys*. 2011; 38:4086–4093. [PubMed: 21859008]
14. Yan X, Pedersen JO, Wei L, Zhang X, Xue R. *IEEE Trans Biomed Eng*. 2015; 62:1652–1659. [PubMed: 25706499]
15. Yan X, Zhang X, Feng B, Ma C, Wei L, Xue R. *IEEE Trans Med Imaging*. 2014; 33:1781–1787. [PubMed: 24710826]
16. Connell IR, Gilbert KM, Abou-Khousa MA, Menon RS. *IEEE Trans Med Imaging*. 2015; 34:825–835. [PubMed: 25838388]
17. Yan X, Zhang X, Wei L, Xue R. *Quant Imaging Med Surg*. 2014; 4:79–86. [PubMed: 24834419]
18. Yan X, Zhang X, Wei L, Xue R. *Appl Magn Reson*. 2015; 46:59–66.
19. Kozlov M, Turner R. *J Magn Reson*. 2009; 200:147–152. [PubMed: 19570700]
20. Lemdiasov RA, Obi AA, Ludwig R. *Concept Magn Reson A*. 2011; 38A:133–147.
21. Yan X, Ma C, Shi L, Zhuo Y, Zhou XJ, Wei L, Xue R. *Appl Magn Reson*. 2014; 45:437–449.
22. Kozlov M, Turner R. *Progress In Electromagnetics Research Symposium Proceedings*. 2010; 6:395–399.
23. Hoult DI. *J Magn Reson Imaging*. 2000; 12:46–67. [PubMed: 10931564]
24. Ocali O, Atalar E. *Magn Reson Med*. 1998; 39:462–473. [PubMed: 9498603]
25. Cao Z, Oh S, Sica CT, McGarrity JM, Horan T, Luo W, Collins CM. *Magn Reson Med*. 2014; 72:237–247. [PubMed: 24006153]
26. Kellman P, McVeigh ER. *Magn Reson Med*. 2005; 54:1439–1447. [PubMed: 16261576]
27. Pruessmann KP, Weiger M, Scheidegger MB, Boesiger P. *Magn Reson Med*. 1999; 42:952–962. [PubMed: 10542355]
28. Yan, X.; Zhang, X.; Ma, C.; Wei, L.; Xue, R. *Proceedings 22th Scientific Meeting; International Society for Magnetic Resonance in Medicine*; 2014. p. 1322

Figure 1. Modeled coils in the EM simulation. (A) An 8-channel ICE-decoupled loop array. (B) An 8-channel capacitively decoupled loop array. (C) A single loop.

Figure 2. Simulated S-parameter plots of the 8-channel ICE-decoupled loop array (A) and capacitively decoupled loop array (B). The reflection coefficient of each loop element was better than -30 dB, indicating all loop elements were well matched to 50 ohm.

Figure 3. Normalized noise covariance matrixes of the 8-channel capacitively-decoupled (A) and ICE-decoupled loop array (B).

Figure 4. Simulated B_1 field (A–C) and H field (D–F) of one coil element of the single loop, the 8-channel capacitively decoupled array and the 8-channel ICE-decoupled array. For the 8-channel array, one coil element was excited with 1 W power and all the other coil elements were terminated with 50Ω .

Figure 5.
Current distribution of the 8-channel ICE-decoupled loop array.

Figure 6. Normalized SNR maps of the 8-channel ICE-decoupled array (**A**) and the capacitively decoupled array (**B**). The SNR was calculated by Kellman's method [26]. Compared with capacitively decoupled array, ICE-decoupled array has higher SNR at the peripheral areas and slightly lower SNR at the central area, which is consistent with the experimental results in our previous work [15].

Figure 7.

G-factor maps of the 8-channel ICE-decoupled array (**top row**) and the capacitively decoupled array (**bottom row**) with R varied from 2 to 4. Compared with capacitively decoupled array, the ICE-decoupled array exhibit lower g-factors and thus better parallel imaging performance, which is consistent with the S-parameter and noise covariance results and the experimental results [15].