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Optimizing the ICE decoupling element distance to improve monopole antenna arrays for 7 Tesla MRI

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Abstract

The induced current elimination (ICE) method has been previously applied to decouple monopole coil arrays in ultrahigh field MRI. However, the method creates low B_1^+ spots near the decoupling elements. In this study, we aim to improve the performance of ICE-decoupled monopole array in human head imaging at 7 Tesla. Eight-channel ICE-decoupled monopole arrays were optimized by varying the position of the decoupling elements. A series of numerical studies were performed using the co-simulation method. In simulation, decoupling performance, quality (Q-) values and transmit field (B_1^+) were comparatively investigated. In addition, we constructed an optimized ICE-decoupled monopole array and compared its performance with the unoptimized array. The simulation results showed that a good trade-off between decoupling and B_1^+ loss can be obtained when decoupling elements were moved 2.5-cm away from coil elements. This was validated by in-vivo MR imaging using the constructed array. Compared with the unoptimized ICE decoupled monopole array, the optimized array had a more homogeneous transmit field and no dark spots or signal cancellations in the MR images.

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Informed consent: Informed consent was obtained from the volunteer included in the study.

Keywords

ICE; magnetic wall; decoupling; homogeneous transmit field; monopole; ultrahigh field MRI

Introduction

Radiative coils such as dipole and monopole antennas have been proposed for ultrahigh field MRI to obtain stronger transmit fields and higher signal-to-noise ratio (SNR) deep into human tissues [1–5]. Due to their unique coil structure, conventional decoupling methods such as element overlapping [6], transformers [7,8] and traditional L/C networks [9] are not easy to fabricate for radiative arrays. The induced current elimination (ICE) or magnetic wall decoupling method [10,11], which does not involve physical connection between coil elements and decoupling circuits, has proved to be a simple and efficient approach for decoupling monopole and dipole arrays in MR imaging [12,13].

The ICE method utilizes one or more physically independent resonators or antennas, referred to as the decoupling elements, to reduce the electromagnetic (EM) coupling among coil elements. To fully compensate the induced current caused by EM coupling, the current along the decoupling element is adjusted by changing its self-impedance [10]. The induced current along the decoupling element usually flows in an opposite direction compared to the current along coil elements. This generates a shielding effect and acts like a “magnetic wall”, which makes the magnetic field of each coil element more locally focused and directed away from nearby elements.

The shielding effect was shown to improve the reception of MR signals in terms of locally enhanced SNR and improved parallel imaging performance [14,15]. However, the shielding effect also disturbs the original EM field. For the ICE-decoupled microstrip line arrays and loop arrays, the decoupling element is a near-field resonator (microstrip or loop resonator) itself and has a very small dimension. Therefore the decoupling element has little influence on the original B_1 field. For the ICE-decoupled monopole array, however, the decoupling element is also a kind of radiative antenna and thus has a more significant shielding effect. That effect significantly decreases the B_1 field of the ICE-decoupled monopole array near the decoupling elements, ultimately resulting in dark spots in the MR images [16].

Assuming decoupling elements are moved further away from the imaging subject, their shielding effect is expected to be reduced and the dark spots should be reduced, at a cost of reduced decoupling performance. In this study, we aim to optimize an 8-channel ICE-decoupled monopole array by varying the positions of decoupling elements. The optimization was conducted based on full-wave EM simulations. An optimized ICE-decoupled monopole array was then constructed and its imaging performance was compared with an unoptimized ICE-decoupled monopole array in bench tests and MRI experiments.

Materials and Methods

EM simulations were conducted using the commercial ANSYS HFSS software (Canonsburg, PA, USA). Figure 1A shows a cross-section of the 8-channel head ICE-

decoupled monopole array. Eight monopole elements were equally spaced along the surface of a cylindrical former (25 cm in diameter). Eight decoupling elements were distributed between the monopole elements. The distance between the decoupling elements and the coil former (D_d) was varied from 0 cm to 3 cm with a step of 0.5 cm. $D_d=0$ corresponds to the conventional unoptimized ICE-decoupled array and was used for the baseline comparison. Both the monopole elements and decoupling elements were made of 10-mm-wide copper tape with 25 cm length.

A cylindrical water phantom (length 37 cm, diameter 16 cm) was placed in the center of the monopole coil arrays, as shown in Figure 1B. The EM parameters of the phantom were: conductivity $\sigma = 0.59$ S/m; relative permittivity $\epsilon_r = 78$. The distance between the boundary and monopole elements was larger than $\lambda/2$. A manual mesh was used to accelerate simulation convergence and the convergence condition S was set to 0.02 to achieve more reliable results. Values of all capacitors were obtained using the circuit co-simulation method [17].

Results

a) Simulation results

Figure 2 plots the reflection coefficient (S_{11}) of each monopole element and the transmission coefficients between adjacent elements (S_{21}), next adjacent elements (S_{31}) and opposite elements (S_{51}). The monopole elements of all arrays were well-matched to 50 Ohms at 297.2 MHz, with S_{11} better than -30 dB, as shown in Figure 2A. This ensures the reliability of the quantitative comparison.

S_{21} plots of monopole arrays with different D_d are shown Figure 2B. The plots show that isolation between adjacent elements decreases as D_d increases as expected. For a distance of $D_d=2.5$ cm, the S_{21} (about -15.1 dB) at 297.2 MHz is acceptable for a 7T MRI transceiver array [18,19]. But the isolation decreased to -12 dB when D_d increased to 3 cm. Figure 2C shows the S_{31} plots of the monopole arrays with different D_d . The isolation between the two next adjacent elements varied little for all cases, from -22 dB to -24.3 dB. The isolation between opposite elements was -22 dB even at the maximum $D_d=3$ cm, as shown in Figure 2D. These simulated S-parameter results show that the D_d should be smaller than 3 cm to achieve excellent decoupling performance between any two elements.

It should be noted from Figure 2A that the elements' Q values decrease as the distance D_d increases. It is known that a single monopole coil has a very low Q value (~ 3) when loaded. Therefore ICE-decoupled monopole arrays with larger D_d have more similar Q values to that of a single monopole.

Figure 3 shows the B_1^+ fields of ICE-decoupled monopole arrays with different D_d . The B_1^+ is denoted as the transmit magnetic field can be extracted from the simulations by [20]:

$$B_1^+ = \frac{B_x + iB_y}{2}.$$

These arrays were excited in birdcage-like mode, applying 1 W power to each port, with sequential 45 degree phase increments. For the unoptimized array with $D_d=0$ cm, the B_1^+ field is low near the decoupling elements, as indicated by the white circles in Figure 3A. As D_d increases, this B_1^+ decrease becomes less significant. It can be concluded from Figure 3 that the dark spots in human images are mainly caused by diminished B_1^+ in these areas.

b) Experimental validation

To validate the simulation results, ICE-decoupled monopole arrays without ($D_d=0$ cm) and with optimization ($D_d=2.5$ cm) were built for comparison, as shown in Figure 4. S-parameter plots of both arrays (loaded with the water phantom) were measured by an Agilent 5071C network analyzer. Gradient-recalled echo (GRE) images and transverse B_1^+ maps were also acquired for comparison. The imaging acquisition parameters were: flip angle (FA)=25 degree, TR=120 ms, TE=6 ms, FOV=250×250 mm², matrix=256×256, slice thickness=5 mm, bandwidth=260 Hz/pixel. B_1^+ profiles on the same human head were mapped with a Turbo FLASH method [21] and normalized in flip angle. SNR comparison in RSS (root-sum-of-squares) combined GRE images were also shown. The SNR was determined by: $SNR=SI/SD \times 0.655$, where SI is the signal intensity and SD is the standard deviation of the noise. The signals were measured from a square of 15×15 pixels in each of the five positions at the center and periphery of the images. The standard deviation was calculated from a 50×50 square in the image background.

Human MR imaging experiments were performed on a whole-body MRI scanner (7T MAGNETOM, Siemens Healthcare, Erlangen, Germany). A healthy female volunteer was scanned subsequently with the unoptimized and optimized arrays (with written informed consent). The human MRI experimental protocol was approved by the local Institutional Review Board (IRB).

The measured S_{21} between adjacent elements is about -15 dB for optimized array and about -25 dB for unoptimized array, which is consistent with the simulation results. The average loaded Q values of each monopole element of the optimized and unoptimized arrays were approximately 16 and 33, respectively.

Figure 5 shows B_1^+ maps and GRE images in the human head using the optimized array and the unoptimized array. For unoptimized ICE-decoupled monopole array, dark spots were found at the peripheral areas in both B_1^+ profile and MR image, as shown in Figs. 5B and 5D. For optimized array, however, the B_1^+ field and MR image were more homogeneous and no dark spots were observed. The optimized array also has more uniform SNR (higher in the middle of the brain and lower at the surface). This result is consistent with the measured Q values.

Discussion

The induced currents in ICE decoupling elements flow in an opposite direction compared to that of coil elements. Therefore the decoupling element acts like a “magnetic wall” and provides a shielding effect. This decreases B_1 field intensity near the decoupling elements, and increases it near the coil elements. Unlike microstrip and loop arrays, the shielding

effect of ICE-decoupled monopole array is much more significant and produces dark spots in the periphery of MR images.

To reduce the shielding effect and avoid the dark spots in MR images, in this study we moved the decoupling elements further away from the coil array and the imaged object while keeping the monopole element positions fixed. However, sufficient isolation between coil elements cannot be attained if the distance between decoupling elements and coil elements is too large. Thus there is a tradeoff between the decoupling performance and the shielding effect. In this study, we optimized the position of decoupling elements based on full-wave EM simulations.

The simulation results demonstrated that for the described 7T 8-channel monopole array, $D_d = 2.5\text{cm}$ was a good tradeoff to balance decoupling performance (better than -15 dB) with reduced shielding effect. This was further validated by experimental results in the human head. Compared with the unoptimized ICE-decoupled monopole array, the optimized array had no dark spots in human head transmit field and GRE images. We note that the optimized ICE-decoupled monopole array had higher transmit efficiency and better SNR in the center of the head compared to the unoptimized array, and both were overall more uniform.

From the simulation results we found that the Q values (Figure 2) varied with different D_d . These simulation results were also validated by experimental results. This finding is also in agreement with our previous work [16]. For $D_d = 0$, the monopole elements have high Q values ($Q_{unloaded}/Q_{loaded} = 110/33$) [16] and high signal intensity at the peripheral areas. When no decoupling methods was applied, which can be assumed as $D_d = \infty$, the monopole array has a very low Q value ($Q_{unloaded}/Q_{loaded} = 9.3/3.1$) [3] and high signal intensity in the middle of the brain. This can be understood by considering that the decoupling elements manipulate the B_1 field of monopole array to a certain degree. Therefore ICE-decoupled monopole arrays with suitable D_d have the potential to provide very homogeneous MR images at 7T.

Based on our theoretical analysis, any kind of resonator can be used as the decoupling element as long as it meets the conditions concluded in the literature [10]. Therefore another possibility to optimize the ICE-decoupled monopole array would be to use L/C loop or microstrip resonators (one or more resonators) [22,23] as decoupling elements.

Conclusion

In summary, we improved the performance of an 8-channel ICE-decoupled monopole array by optimizing the distance of the decoupling elements from the array. The optimal distance for our array configuration was found to be 2.5 cm. Compared with an unoptimized ICE decoupled monopole array, the optimized array had more homogeneous transmit field and no dark spots or signal cancellations in the MR images.

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Figure 1.

Cross-sectional array diagram (A) and simulation model (B) of the 8-channel ICE-decoupled monopole array with different D_d . Note that D_d is the distance between the decoupling elements and the coil former.

Figure 2.

Simulated S-parameter plots of 8-channel ICE-decoupled monopole arrays with varying decoupling element-to-array distance D_d , where D_d ranges from 0 cm to 3 cm with a step of 0.5 cm.

Figure 3.

Simulated B_1^+ maps of 8-channel ICE-decoupled monopole arrays with different D_d . These arrays were excited in birdcage-like mode, applying 1 W power to each port, with a sequential 45 degree phase increment.

Figure 4.
Photographs of 8-channel ICE-decoupled monopole arrays without (A: $D_d=0$ cm) and with (B: $D_d=2.5$ cm) optimized decoupling element-to-array distances.

Figure 5. Measured B_1^+ maps and GRE images in a volunteer's brain for the 8-channel ICE-decoupled monopole arrays with (A and C: $D_d = 2.5$ cm) and without optimization (B and D: $D_d = 0$ cm). The dark spots of the ICE-decoupled monopole array (white arrows) are diminished for the optimized array. SNR is also reported for the GRE images, and is more uniform in the optimized case.