RF Shimming Considering Coupling Effects for High-Field MRI

Hye Young Heo, Min Hyoung Cho, Soo Yeol Lee

Department of Biomedical Engineering, Kyung Hee University, Korea

(Received April 8, 2008. Accepted June 23, 2008)

Abstract

The RF shimming technique has been used to improve the transmit RF field homogeneity in highfield MRI. In the RF shimming technique, the amplitude and phase of the driving currents in each coil element are optimized to get homogenous flip angle or uniform image intensity. The inductive and capacitive coupling between the coil elements may degrade the RF field homogeneity if not taken into account in the optimization procedure. In this paper, we have analyzed the coupling effects on the RF shimming using a sixteen-element TEM RF coil model operating at 300 MHz. We have found that the coupling effects on the RF shimming can be reduced by putting high dielectric material between the active rung and the shield.

Key words: MRI, high field, B_1 homogeneity, transmit array coil, RF shimming

I. INTRODUCTION

As the main field strength of magnetic resonance imaging (MRI) increases above 3.0 Tesla, the wavelength of the radiofrequency (RF or B_1) field for MRI becomes comparable to the dimension of the coil itself and the human body. The shortened wavelength might produce significant standing waves in the human body, which in turn degrades the RF field homogeneity. Due to the shortened wavelength, the B_1 homogeneity of the high field RF coil is much poorer than the low field RF coil. Furthermore, electromagnetic field attenuation in the human body is accentuated at high field strength, which degrades the B_1 homogeneity further [1-2]. Recently, a number of methods to improve the B_1 homogeneity in high field MRI have been proposed. In most methods, phase-array transmit RF coils are used to improve the B_1 homogeneity since we have more degrees of freedom to change the RF field distribution by adjusting the RF currents at each coil element [3-6]. It is believed that more coil elements are necessary at higher fields to secure acceptable B_1 homogeneity. Recent simulation studies suggest that at least 16 elements are necessary at 7.0 T and 80 elements at 14.0 T to secure the B_1 homogeneity using the RF shimming technique [7-8]. However, when more elements are used to shim the RF field, the spacing between the coil elements becomes so small that the inductive and capacitive couplings between the neighboring coil elements become significant. In the previous studies, the coupling effects on the RF shimming have been neglected given that the couplings could be removed somehow in the RF coil fabrication. But, it is considered that the couplings cannot be completely removed in practical RF coil developments.

In this study, we have analyzed the coupling effects on the RF shimming at 300MHz, and we propose a technique to reduce the coupling effects. Computer simulation results obtained with the FDTD method are presented.

Corresponding Author : Soo Yeol Lee, Professor
Dept. of Biomedical Engineering Kyung Hee University, Korea
Tel : +82-31-201-2980, Fax : +82-31-201-3666
E-mail : sylee01@khu.ac.kr

This study was supported by a grant from the Korea Ministry of Health and Welfare (02-PJ3-PG6-EV07-0002) anda grant from the Korea Science and Engineering Foundation (R11-2002-103).

II. METHODS

A. RF Coil Modeling

The 16-element TEM head coil has been used together with the head model consisting of 23 different tissues as shown in Fig. 1. The coil and head model consist of rectangular elements with the isotropic resolution of 5mm. To improve the RF coil efficiency in human brain imaging, we used the elliptical coil structure having two layers. The active rungs are placed on the inner ellipse having the major and minor axes of 24cm and 21cm, respectively. The active rung is 2cm wide and

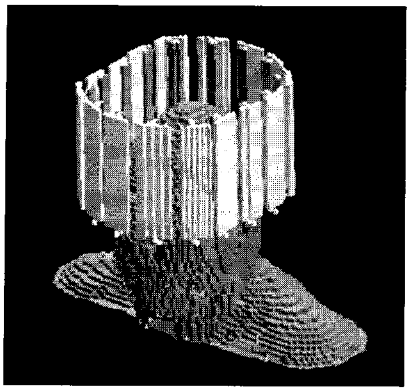

Fig. 1. The 16-element RF coil and human head model

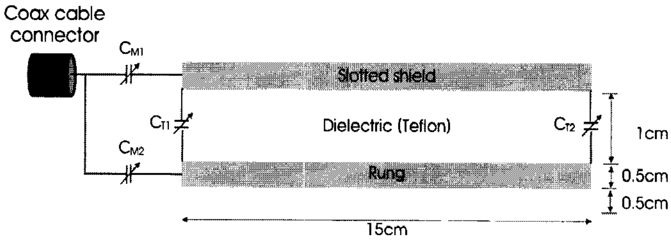

Fig. 2. A schematic diagram of the coil element having the Teflon layer

15cm long and it is made of copper. We chose rather short rung length since high field RF coils are usually targeted to human brain imaging. On the inner elliptical surface, the active rungs are positioned vertically based on the equal spacing criterion. Both ends of each rung are connected to the slotted shield made of copper. The slotted shields are placed on the outer ellipse having the major and minor axes of 27cm and 23cm, respectively. The slotted shield length is 15 cm and the slot width is 0.5 cm. To feed RF currents to an element, we used two current sources at both ends of the element simulating the balanced driving mode. The current sources at both ends have the same magnitude but opposite phase. Each element was individually tuned at 300MHz and matched to the nominal 50 ohm line.

To evaluate the RF field homogeneity in the human head model, we constructed xFDTD models for the human head placed in the 16-element TEM RF coil. To evaluate the coupling effects on the B_1 homogeneity, we have made three RF coil models as below:

1) Model 1 (no coupling): Each coil element has been tuned at 300 MHz individually by adjusting the tuning capacitors. When a coil element is driven by a Gaussian current source, all other elements are intentionally deactivated by disconnecting the tuning capacitors between the rung and the shield.
2) Model 2 (ordinary coupling): The coil elements are the same as in the case of Model 1. But, all the coils are active, that is the tuning capacitors are not disconnected, when one of the elements is driven by the Gaussian current source.
3) Model 3 (reduced coupling): To reduce the coupling between the neighboring elements in Model 2, a high dielectric Teflon layer was placed between the rung and the shield as shown in Fig. 2. Due to its high dielectric constant, Teflon effectively doubles the electrical distance between elements as compared to the air. Two capacitors, C_{M1}, C_{M2}, are used to match the coil to the nominal 50 ohm line and C_{T1} and C_{T2} are used to tune the resonance frequency at 300MHz.

B. RF Shimming

We first calculated the RF field maps produced by each element of the RF coil models using the xFDTD tool (Remcom Inc., College station, PA, USA). In the calculation of the RF field maps, we used identical current sources, in terms of magnitude and phase, for all the coil elements. With the calculated field maps, we calculated optimal current magnitudes and phases at each element to produce homogeneous RF field in the region of interest (ROI). In the optimization using the conjugate gradient method, the optimal current magnitudes and phases were searched to minimize the error function all over the ROI in the human model. The final field distribution was obtained by adding up the field maps of every element after weighting them by the optimal magnitude and phase. The circularly polarized clockwise field, B_1^+, at the ROI in the head model due to the n-th element has x- and y-components. The x- component is denoted as:

$$B_{1xn} = A_n e^{j(wt + \varnothing_n)} a_{ns} e^{jb_n x} \tag{1}$$

where A_n and $\varnothing_n$ represent the weighting factors for the current magnitude and phase at the n-th element, respectively, w is the Larmor frequency. The term $a_{nx} e^{jb_n x}$ is the x-directional RF field map of the n-th element when it is driven by the current source with unit amplitude and zero phases. A similar equation can be written for the y-component of the B_1^+ field. The final x-and y-directional B_1^+ sensitivity of the circularly polarized clockwise component were then calculated by summing up all B_{1xn} and B_{1yn}, respectively, of all the elements.

To optimize the B_1 homogeneity in the ROI, cost functions were defined to minimize the relative standard deviation (RSD) of the flip angle or the gradient echo signal intensity (SI_{GE}) over the ROI. The flip angle at a given point was calculated as below:

$$\alpha = \gamma\tau\left|\sum_{n=1}^{N} B_{1n}^{+}\right| \tag{2}$$

where γ is the gyromagnetic ratio, τ is the RF pulse duration, N is the number of elements, and B_{1n}^{+} is the circularly polarized clockwise component produced by the n-th element. The gradient echo signal intensity is calculated by:

$$SI_{GE}=\left|\sin\left(\gamma\tau\left|\sum_{n=1}^{N} B_{1n}^{+}\right|\right)\right|. \tag{3}$$

To reduce the probability of the optimization to be trapped in a local minimum, each optimization has been performed several times with different initial phase and amplitude settings. The initial phase values were between 0 and 2π and the initial amplitudes were between 0 and 1. The optimizations have been performed using the MATLAB functions on a 1.6GHz workstation having two quad CPUs and 16 GB random access memory.

Ⅲ. RESULTS

On the left hand side of Fig. 3a, 3b and 3c, we illustrate the magnitude of B_1^+ fields of each element obtained without coupling (Model 1), with coupling (Model 2) and with reduced coupling (Model 3), respectively. In Model 3, 10mm thick Teflon layer with the relative dielectric constant of 2 was used to reduce the couplings between the elements. Only four field maps are illustrated in the figure without any preference. In the right hand side of the figures, we show the total B_1^+ field maps which have been obtained by simply adding up all the fields produced by the elements without optimizing the RF currents.

The total RF field in Fig. 3a shows the typical high B_1^+ field at the central region due to the standing wave effect at

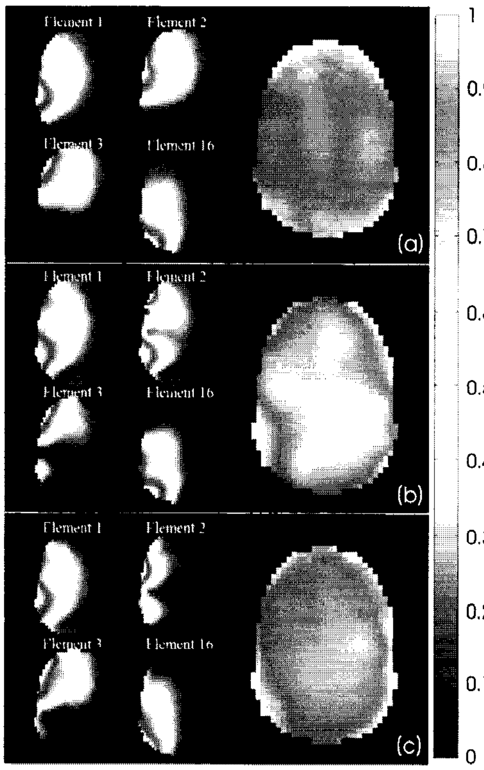

Fig. 3. The B_1^+ fields at the central plane of the head model at 300MHz. (a) B_1^+ field in Model 1. (b) B_1^+ field in Model 2. (c) B_1^+ field in Model 3.

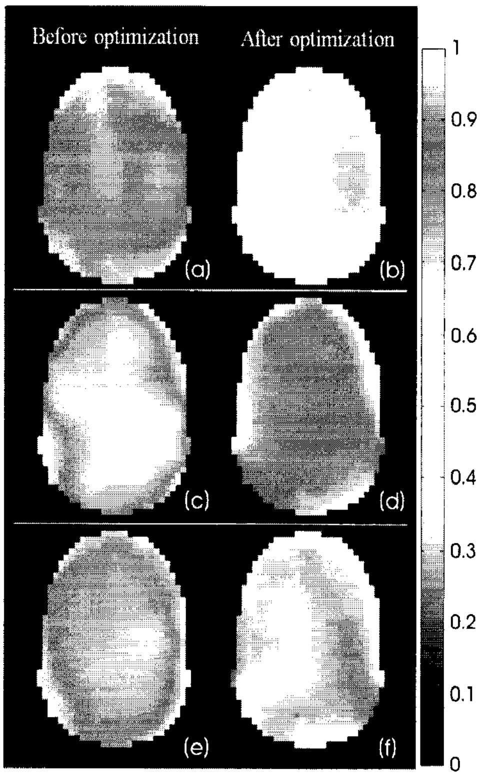

Fig. 4. The B_1^+ fields map before the shimming (left images) and after the shimming (right images). (a) (b) B_1^+ field in Model 1. (c) (d) B_1^+ field in Model 2. (e) (f) B_1^+ field in Model 3.

Table 1. Relative standard deviation of the flip angle and the signal intensity.

	Optimize α		Optimize SI_{GE}	
	RSD before shim	RSD after shim	RSD before shim	RSD after shim
Model 1	0.12	0.02	0.05	0.016
Model 2	0.33	0.07	0.1	0.04
Model 3	0.26	0.05	0.07	0.03

300MHz. Contrary to Fig. 3a, the RF field map of each element in Fig. 3b has wider sensitive region due to the coupling. The coupling effects induce RF currents in other elements when one element is fed by RF current. The total field map in Fig. 3b is significantly different from Fig. 3a implying that the coupling may have significant effects on the RF shimming. In Fig. 3c, we can notice that the sensitive region of each element has been narrowed due to the coupling reduction. The resulting total B_1^+ field in Fig. 3c is more uniform and symmetric than Fig. 3b. This result suggests that the RF field homogeneity can be improved by reducing the couplings between the elements.

Figure 4 shows the RF shimming results for the three models at 300MHz. The left hand and right hand side images show the total RF field obtained before and after the optimization, respectively. The optimizations were performed with respect to the flip angle on the axial slice of the head model without any preference. The optimizations with respect to the gradient echo signal intensity also resulted in very similar RF field patterns. As can be noticed from Fig. 4b, the RF uniformity is much improved by the RF shimming when there is no coupling between the elements. Figure 4d shows that the coupling effect degrades the RF field uniformity even after the RF shimming. With reducing the coupling using the high dielectric layer, the RF uniformity can be partially recovered as can be seen in Fig. 4f. Table 1 summarizes the RSDs for all the cases. As compared to the case of Model, the RSDs in Model 2 are much increased in both flip angle and the signal intensity. This indicates that the RF shimming may have significant errors if the coupling effect is not considered in the optimization process. In the case of Model 3, the RSDs are in between the ones of Model 1 and Model 2.

The optimized magnitudes and phases for the three models are depicted in a polar plot in Figure 5. The indices around the outer circles indicate the coil element number. In the magnitude plots, the radii mean the relative current magnitudes with respect to the ideal case in which all the current magnitudes are 1. In the phase plots, the radii mean the phase deviations from the ideal case in which the phase at each element increases by 22.5 degrees along the azimuthal direction.

Ⅳ. DISCUSSIONS AND CONCLUSIONS

The B_1 field inhomogeneity is troublesome in high field MRI over 3.0T. Various methods have been proposed to mitigate the B_1 field inhomogeneity effect. The RF shimming using multi-channel transmit array coils are among them. It has been suggested that more coil elements are necessary at higher fields to secure acceptable B_1 homogeneity because the larger number of adjustable factors [7-8]. However, the possible couplings between the coil elements have not been considered in the RF shimming procedures. If we increase the number of coil elements, the spacing between the elements becomes so close that the coupling effects jeopardize the RF shimming. In this study, we considered the 16-element RF coil. We think that the coupling effects are much bigger in the 32- or 64-element RF coils. We have found that the coupling effects on the RF shimming can be significantly reduced by employing high dielectric layers between the active rung and the shields. The RF fields are focused in the dielectric layers,

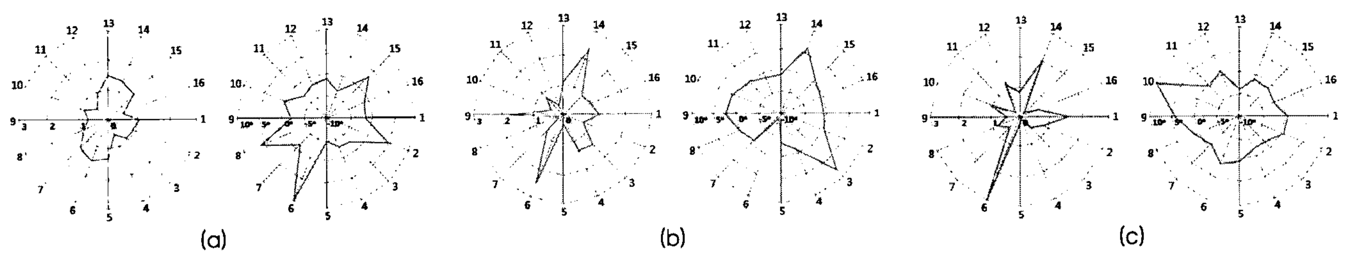

Fig. 5. The optimized amplitudes (left) and phases (right) on the polar coordinates. (a) Model 1 (b) Model 2 (c) Model 3

hence, the coupling effects are reduced. The couplings may be reduced in many other ways. It has been recently reported that decoupling capacitors connecting the middle of the active rungs reduce the couplings significantly in the TEM coil operating at 300MHz [6].

In this study, we considered only the central axial slice in the RF shimming without any preference. It has been reported that the RF field uniformity at other slices than the optimized slice might be degraded as compared to the case the RF shimming is not used. Therefore, the coupling effects should be investigated further in the RF shimming at other slices or in the whole brain.

For the practical use of the RF shimming technique in the clinical applications, the RF field map of every coil element should be measured fast. Since the RF field maps should be measured element by element in the whole 3D space, fast imaging sequences are essential for the RF shimming. Any errors in measuring the RF field map would degrade the RF shimming performance significantly. It is thought that efficient and accurate RF field map measuring techniques should be developed for practical use of the RF shimming technique. Parallel excitation hardware, multiple RF power amplifiers and couplers, are other big issues for practical use of the RF shimming.

In conclusion, the couplings between the elements of the high frequency RF coil significantly degrade the RF shimming efficacy in high field MRI. For practical implementation of the RF shimming, the couplings should be reduced in the RF coil design and the coupling effects should be considered as well in the RF shimming process. It is expected that the coupling effects can be reduced significantly by using high dielectric layers between the active rungs and the shields.

REFERENCES

[1] Bottomley P, Andrews E. "RF magnetic field penetration, phase shift and power dissipation in biological tissue: implications for NMR imaging", *Phys Med Biol* 1978;23:630-643

[2] Glover GH, Hayes CE, Pelc NJ, Edelstein WA, Mueller OM, Hart HR, Hardy CJ, Odonnell M, Barber WD. "Comparison of linear and circular-polarization for magnetic-resonance imaging", *J Magn Reson* 1985;64:255-270

[3] Zhu Y, Watkins R, Giaquinto R, Hardy C, Kenwood G, Mathias S, Valent T, Denzin M, Hopkins J, Peterson W, Mock B. "Parallel excitation on an eight transmit-channel MRI system", *Proceedings of the 13th Annual Meeting of the ISMRM, Miami Beach, FL, USA*, 2005. P 14.

[4] Adriany G, Van de Moortele PF, Wiesinger F, Moeller S, Strupp JP, Andersen P, Synder C, Zhang X, Chen W. Pruessmann KP, Boesiger P, Vaughan T, Ugurbil K. "Transmit and receive transmission line arrays for 7 Tesla parallel imaging", *Magn Reson Med* 2005;53:433-445

[5] Hoult DI, Kolansky G, Kripiakevich D, King SB. "The NMR multi-transmit phased array: a Cartesian feedback approach", *J Magn Reson* 2004;171:64-70

[6] Collins CM, Liu W, Swift BJ, Smith MB. "Combination of optimized transmit arrays and some receive array reconstruction methods can yield homogeneous images at very high frequencies" *Magn Reson Med* 2005;54:1327-1332

[7] Mao W, Smith MB, Collins CM, "Exploring the limits of RF shimming for high-field MRI of the human head", *Magn Reson Med* 2006;56:918-922

[8] Collins CM, Wang Z, Mao W, Fang J, Liu W, Smith MB. "Array-optimized composite pulse for excellent whole brain homogeneity in high-field MRI", *Magn Reson Med* 2007;57: 470-474