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Electromagnetic Field and Radio Frequency Circuit Co-Simulation for Magnetic Resonance Imaging Dual-Tuned Radio Frequency Coils

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Abstract

Electromagnetic (EM) field simulation plays a key role in the design of magnetic resonance imaging radio frequency (RF) coils. However, the values of the components in tuning and matching circuits often need to be iterated repeatedly in the conventional simulation method in order to achieve optimal scattering parameters. This leads to a time-consuming optimization to tune and match RF coils, particularly dual-tuned coils that are comprised of multiple lumped elements. A method combining EM field simulation and circuit simulation was employed in this paper, which can dramatically improve simulation efficiency for the optimization of the values of the lumped elements and corresponding EM field distribution. A systematical theoretical analysis of the co-simulation method was presented. To validate the accuracy and efficiency of the co-simulation method, a comparison study was conducted between the conventional simulation and the proposed co-simulation approaches.

Index Terms

Computational electromagnetics (EMs); magnetic resonance imaging (MRI); numerical analysis; radio frequency (RF) coil

I. Introduction

Magnetic resonance imaging (MRI) is one of the primary imaging modalities used in current biomedical research and clinical applications. In the MRI systems, radio frequency (RF) coils are critical components used to transmit and receive magnetic resonance signals [1]–[4]. In order to improve the sensitivity of signal detection and to obtain high-quality MRIs,

the performance of RF coils needs to be optimized. To design high-performance RF coils, numerical electromagnetic (EM) simulation is often employed as a cost-efficient design approach and a theoretical method to study the EM interactions between the load and RF coils [5]. Through the simulation, the EM field and specific absorption rate (SAR) distribution can be investigated to ensure imaging quality and patient safety [6].

The conventional EM numerical analysis method is primarily based on the finite element method and finite-difference time-domain algorithms [7]–[9]. In these approaches, the coil structure with tuning and matching circuits are built directly into the model. The scattering parameters, EM field, and SAR distribution can be obtained to evaluate the coil performance [10], [11]. However, the values of lumped elements in the tuning and matching circuits of the RF coils need to be optimized. The optimization is time-consuming since the field distribution must be re-calculated with each iteration. The optimization is inefficient and impractical, particularly in the case of a complex circuit with multiple components such as dual-tuned coils.

The EM field and RF circuit co-simulation approach simulates the field distribution and tuning/matching conditions of the RF coils separately [12]. In this paper, an RF coil co-simulation strategy that can dramatically increase calculation speed and shorten the computation time is demonstrated. The theoretical analysis of the co-simulation method is proposed. It is subsequently utilized to optimize with high efficiency the tuning and matching circuit of a dual-tuned coil array and make the numerical calculation practical, particularly in complex RF circuits. The accuracy of the co-simulation method is validated by comparing scattering parameters and field distribution results with the conventional simulation method.

II. Theory

Compared with the conventional simulation method, tuning of the lumped elements is conducted in the circuit simulation in the co-simulation approach, which means that the field distribution does not need to be repeatedly calculated during the iteration. When the RF coil is well tuned and matched, the flowchart of the two simulations is shown in Fig. 1. There are linear assumptions in the method with a few limits. The proposed method applies to calculate the indicators when the linearity assumption is applicable such as the S-parameters and the EM fields. However, the SAR needs to be further calculated. The EM field and SAR are determined by the coil structure, the tuning and matching circuit, and the input power of each coil which can be expressed as

$$Y^i(\omega) = H_{\text{field}}^i(\omega) \cdot H_{\text{circuit}}^i(\omega) \cdot X_{\text{circuit}}^i(\omega) \quad (1)$$

where i and ω denote the port index and angular frequency, respectively. $Y^i(\omega)$ represents the EM field, $H_{\text{field}}^i(\omega)$ signifies the transfer function of field distribution (calculated in the EM field simulation), and $H_{\text{circuit}}^i(\omega)$ and $X_{\text{circuit}}^i(\omega)$ denote the transfer function of the

tuning/matching circuit (calculated in the circuit simulation) and the input power of each excitation port of the circuit, respectively.

$H_{\text{field}}^i(\omega)$ can be calculated by using the Fourier trans-form of the EM field under the excitation of the impulse function $\delta(t)$. This can be defined as

$$H_{\text{field}}^i(\omega) = F[\text{EM}(t)] = \int_0^{+\infty} \text{EM}(t) e^{-j\omega t} dt = \lim_{\substack{\Delta t \rightarrow 0 \\ Nt \rightarrow \infty}} \sum_{n=0}^{Nt} \text{EM}(n\Delta t) e^{-j\omega n\Delta t} \Delta t \quad (2)$$

where F denotes the Fourier transform of the EM field data and $\text{EM}(t)$ represents the EM field. t denotes the time step width, and n and Nt denote the number and the max number of time steps in the simulation, respectively.

An N-channel coil array can be modeled as an N-port network, as shown in Fig. 2. This relation represents a linear system of equations for the determination of the unknown scattering matrix S^i of the N-port [13]. The matrices S^i of each subnetwork represent the tuning and matching circuit. The power of the incident wave and the reflected wave of each port are $a_1^i, a_2^i, b_1^i, b_2^i, i = 1, 2, 3, \dots, N$, and are expressed as

$$\begin{bmatrix} a_2^i \\ b_2^i \end{bmatrix} = \begin{bmatrix} -\frac{s_{11}}{s_{22}} & \frac{1}{s_{12}} \\ \frac{s_{12} \cdot s_{21} - s_{11} \cdot s_{22}}{s_{12}} & \frac{s_{22}}{s_{12}} \end{bmatrix} \cdot \begin{bmatrix} a_1^i \\ b_1^i \end{bmatrix} \quad (3)$$

in which the s-parameter matrix of each subnetwork is transferred from the impedance matrix of the equivalent circuit model in the subnetworks. The scattering matrix S^i of the i th subnetwork is defined as

$$S^i = H_{\text{circuit}}^i(\omega) = \begin{bmatrix} -\frac{s_{11}}{s_{22}} & \frac{1}{s_{12}} \\ \frac{s_{12} \cdot s_{21} - s_{11} \cdot s_{22}}{s_{12}} & \frac{s_{22}}{s_{12}} \end{bmatrix} \quad i = 1, 2, 3, \dots, N. \quad (4)$$

This can be reduced to (5) when the coil array is well decoupled and the interaction between ports is negligible

$$\begin{bmatrix} a_2^1 \\ b_2^1 \\ a_2^2 \\ b_2^2 \\ \vdots \\ a_2^N \\ b_2^N \end{bmatrix} = \begin{bmatrix} S^1 & 0 & \dots & 0 \\ 0 & S^2 & \dots & 0 \\ \vdots & \vdots & \ddots & \vdots \\ 0 & 0 & \dots & S^N \end{bmatrix} \cdot X_{\text{circuit}}^i \quad (5)$$

where $X_{\text{circuit}}^i(\omega) = [a_1^1 b_1^1 a_1^2 b_1^2 \dots a_1^N b_1^N]^T$. Since $a_2^i = b_1^{i'}$ and $b_2^i = a_1^{i'}$, it can be deduced to

$$\begin{bmatrix} a_1^{1'} \\ b_1^{1'} \\ a_1^{2'} \\ b_1^{2'} \\ \vdots \\ a_1^{N'} \\ b_1^{N'} \end{bmatrix} = \begin{bmatrix} \epsilon^1 & 0 & \dots & 0 \\ 0 & \epsilon^2 & \dots & 0 \\ \vdots & \vdots & \ddots & \vdots \\ 0 & 0 & \dots & \epsilon^N \end{bmatrix} \cdot \begin{bmatrix} a_2^1 \\ b_2^1 \\ a_2^2 \\ b_2^2 \\ \vdots \\ a_2^N \\ b_2^N \end{bmatrix} \quad (6)$$

in which

$$\epsilon^i = \begin{bmatrix} 0 & 1 \\ 1 & 0 \end{bmatrix}.$$

Based on the above equations, the EM field distribution are expressed as

$$Y^i(\omega) = H_{\text{field}}^i(\omega) \cdot \begin{bmatrix} \varepsilon^1 & 0 & \dots & 0 \\ 0 & \varepsilon^2 & \dots & 0 \\ \vdots & \vdots & \ddots & \vdots \\ 0 & 0 & \dots & \varepsilon^N \end{bmatrix} \cdot \begin{bmatrix} S^1 & 0 & \dots & 0 \\ 0 & S^2 & \dots & 0 \\ \vdots & \vdots & \ddots & \vdots \\ 0 & 0 & \dots & S^N \end{bmatrix} \cdot \begin{bmatrix} a_1^1 \\ b_1^1 \\ a_1^2 \\ b_1^2 \\ \vdots \\ a_1^N \\ b_1^N \end{bmatrix}. \quad (7)$$

III. Applications

As an example, a dual-tuned, three-channel phased array coil which was comprised of three round loops, each with a 40 mm radius and 3 mm conductor width was used. Commercial software (Computer Simulation Technology, Darmstadt, Germany) was used to validate the proposed simulation method in this paper.

A cubic phantom measuring $200 \times 60 \times 80 \text{ mm}^3$ was placed 5 mm above the RF coil model. The material property of the phantom was set as conductivity = 3.53 S/m, relative permittivity $\varepsilon_r = 74$, and relative permeability $\mu_r = 0.99$. In the conventional simulation, the ports and capacitors were arranged at the gaps of the RF coil model, as shown in Fig. 3(a). The red and blue components in Fig. 3(a) denote the ports and lumped elements, respectively. In the proposed simulation, all the lumped elements were replaced by ports with 50Ω impedance, as shown in Fig. 3(b). A dual-tuned circuit was added to achieve the resonant frequency 115.9 MHz for fluorine (^{19}F) and 123.2 MHz for hydrogen (^1H) at 3 T [14]. The dual-tuned circuit structure is depicted in Fig. 4, in which the red triangles denoted the excitation ports.

Time-domain solver was employed, and the linear equation system solver accuracy was set to -30 dB . The boundaries of the RF coil model were all set to open, and the bandwidth was set to 50–200 MHz. The total number of mesh cells was 645 840.

The RF coil was tuned by adjusting the capacitance or inductance of lumped elements. In the conventional simulation, the coils were tuned by adjusting the lumped elements directly in the model and the simulations were all performed in field to obtain the EM field and SAR. In the co-simulation, the coils were tuned by changing the capacitance in the RF circuit and combining the individual EM field based on the tuned data of the external ports in order to obtain the EM field. The trust domain framework algorithm was applied to optimize the simulation results [15]. This algorithm used a domain accuracy value to control convergence behavior. All the capacitance and inductance parameters were selected to optimize the frequency tuning and impedance matching of the coil. The S-parameters (S_{11} , S_{22} , and S_{33}) with the targets were set as $< -30 \text{ dB}$.

The EM field and the SAR data of the xy plane of the two methods were exported and compared to validate the accuracy of the proposed method. The total time required for the two simulations are defined, respectively, as

$$t_{\text{MWS}} = t_{\text{mws-ch}} \cdot n + t_{\text{mws-ch}} \cdot n \cdot (k - 1) \quad (8)$$

$$t_{\text{co-sim}} = (t_{\text{co-p}} \cdot n_e \cdot n + t_{\text{ds}}) + t_{\text{ds}} \cdot (k - 1) \quad (9)$$

in which t_{MWS} and $t_{\text{mws-ch}}$ denoted the total and individual channel simulation time in the conventional procedure, respectively. $t_{\text{co-sim}}$, $t_{\text{co-p}}$ represented the total time and the time for the individual port of the EM field simulation of the proposed method, respectively. t_{ds} was the time in circuit simulation n was the number of channels, n_e was the number of ports for each channel, and k was the number of simulations of the entire process. In this model $n = 3$ and $n_e = 4$.

The human spine model was also applied as the RF coil load. The electric properties of the tissue used for simulation are listed in Table I. The simulation time of the conventional simulation methods and the proposed co-simulation methods was compared with the same mesh cells in two cases: 5 701 104 and 10 370 850 mesh cells.

IV. Results

For one EM field simulation in this phantom case, the energy convergence time of each port is 124.36 ns in the conventional simulation with three ports resulting in a total time 373.08 ns and it is only 18.71 ns in the proposed co-simulation with 12 ports resulting in a total convergence time 224.52 ns. The proposed method saves about 39.82% of the total energy convergence time in one simulation. A total of 84 iterations for both methods were required to achieve the optimized S-parameters. Compared with the conventional simulation, the proposed co-simulation method saves about 95% of total simulation time because it only needs to calculate $H_{\text{circuit}}^i(\omega)$, whereas the conventional method needs to repeatedly calculate $H_{\text{field}}^i(\omega)$ in each iteration. The total time comparison of the two methods is shown in Table II in the case of the phantom as the RF coil load. When loading with the human spine model, the individual channel simulation time of the conventional simulation methods and the proposed co-simulation methods is 124.95 and 41.25 min with 5 701 104 mesh cells, respectively, with 10 370 850 mesh cells, it is 142.75 and 68.65 min, respectively.

Fig. 5 shows the simulated S-parameter of the dual-tuned RF coil obtained from the conventional and the proposed co-simulation methods. The S-parameters agree well in both methods. The EM field and SAR distribution of the xy plane, which is 5 mm above the surface of phantom, are compared. The maximum deviation of the field distribution is below 4% compared with the conventional simulation; the detailed data of the two methods are shown in Figs. 6 and 7. The EM field distribution maps of each channel at the same position

of 115.9 MHz for ^{19}F and 123.2 MHz for ^1H are shown in Fig. 6, respectively. The SAR distribution maps are plotted in Fig. 7.

V. Conclusion

This paper describes a method for enhancing the simulation efficiency of RF coil design. The co-simulation method provides a more efficient choice particularly for complex RF circuits. Without resonant structure in the co-simulation, the convergence of the EM field simulation is faster than the conventional method. Through the three-channel dual-tuned RF coil simulation, the proposed co-simulation reduces 40% simulation duration in one EM field simulation. Regarding to the optimization of the values of the lumped elements, the co-simulation saves more than 90% of the total time. The EM field and SAR deviation between the proposed co-simulation and the conventional simulation are all less than 4% which demonstrate the accuracy of the proposed method.

The systematical theoretical analysis of the co-simulation is based on the linearity assumption and the negligible interaction between the ports. When the decoupling performance is determined by the structure of RF coils, such as overlapping coil elements, the proposed analysis can be used. The non-diagonal elements in the matrixes in (5) and (6) will be non-zero if the coil array is not well decoupled. The calculation is still within the matrix, which should be still faster than numerical calculation of EM fields.

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References

1. Roemer PB, Edelstein WA, Hayes CE, Souza SP, Mueller OM. The NMR phased array. *Magn Reson Med*. 1990; 16(2):192–225. [PubMed: 2266841]
2. Zhang X, Ugurbil K, Chen W. Microstrip RF surface coil design for extremely high-field MRI and spectroscopy. *Magn Reson Med*. 2001; 46(3):443–450. [PubMed: 11550234]
3. Shou G, Xia L, Liu F, Zhu M, Li Y, Crozier S. MRI coil design using boundary-element method with regularization technique: A numerical calculation study. *IEEE Trans Magn. Apr*; 2010 46(4): 1052–1059.
4. Li Y, Xie Z, Pang Y, Vigneron D, Zhang X. ICE decoupling technique for RF coil array designs. *Med Phys*. 2011; 38(7):4086–4093. [PubMed: 21859008]
5. Yan X, Xie Z, Pedersen JO, Zhang X. Theoretical analysis of magnetic wall decoupling method for radiative antenna arrays in ultrahigh magnetic field MRI. *Concept Magn Reson B*. 2015; 45(4):183–190.
6. Collins CM, Smith MB. Calculations of B1 distribution, SNR, and SAR for a surface coil adjacent to an anatomically-accurate human body model. *Magn Reson Med*. 2001; 45(4):692–699. [PubMed: 11283998]
7. Xin X, Wang D, Han J, Feng Y, Feng Q, Chen W. Numerical optimization of a three-channel radiofrequency coil for open, vertical-field, MR-guided, focused ultrasound surgery using the hybrid

method of moment/finite difference time domain method. NMR Biomed. 2012; 25(7):909–916. [PubMed: 22161891]

8. Guo Y, Jiang X. Simulations of the tent artifacts in magnetic resonance imaging. IEEE Trans Magn. Feb; 2012 48(2):659–662.
9. Chi J, Liu F, Xia L, Shao T, Crozier S. An improved cylindrical FDTD algorithm and its application to field-tissue interaction study in MRI. IEEE Trans Magn. Feb; 2011 47(2):466–470.
10. Wang C, et al. A practical multinuclear transceiver volume coil for in vivo MRI/MRS at 7 T. Magn Reson Med. 2012; 30(1):74–78.
11. Amjad AA, Kamondetdacha R, Kildishev AV, Park SM, Nyenhuis JA. Power deposition inside a phantom for testing of MRI heating. IEEE Trans Magn. Oct; 2005 41(10):4185–4187.
12. Kozlov M, Turner R. Fast MRI coil analysis based on 3-D electromagnetic and RF circuit co-simulation. J Magn Reson. 2009; 200(1):147–152. [PubMed: 19570700]
13. Rolfes I, Schiek B. Multiport Method for the Measurement of the Scattering Parameters of N-Ports. IEEE Trans Microw Theory Tech. Jun; 2005 53(6):1990–1996.
14. Hu L, Hockett FD, Chen J, Zhang L, Caruthers SD, Lanza GM. A generalized strategy for designing $^{19}\text{F}/^1\text{H}$ dual-frequency MRI coil for small animal imaging at 4.7 Tesla. J Magn Reson Imag. 2011; 34(1):245–252.
15. Conn, AR., Gouldand, NIM., Toint, PL. Trust Region Methods (MPS-SIAM Series on Optimization). Philadelphia, PA, USA: SIAM; 2000.

Fig. 1. Flowchart of the two simulations. (a) Conventional simulation. (b) Co-simulation.

Fig. 2.
Wave parameters of an N-port.

Fig. 3.
RF coil model for EM field simulation for (a) conventional simulation and (b) co-simulation.

Fig. 4.

Dual-tuned circuit structure of the coils and the values of the lumped elements were achieved by optimization.

Fig. 5.
S-parameters for conventional simulation and co-simulation.

Fig. 6. EM field distribution of ^{19}F for conventional simulation and co-simulation in the water phantom. (a)–(d) “E” represents the electric field. (e)–(h) “H” represents the magnetic field. “STD” represents the standard deviation.

Fig. 7. SAR distribution maps of ^{19}F and ^1H for conventional simulation and co-simulation in the phantom. “STD” represents the standard deviation.

TABLE I

Electric Properties of the Tissue Used for Simulation

Tissue	Conductivity (S/m)	Relative permittivity
Skin	0.49	72.93
Blood	1.23	76.82
Fat	0.04	6.07
Muscle	0.71	65.97
Spinal cord	0.34	47.27

TABLE II

Detailed Data of the Two Methods

	k	Mesh cells	t_d (min)	Each channel of one simulation time (min)	Total time (min)
Phantom (Conventional)	84	645840	N/A	$t_{miss-d} = 54.15$	13645.8
Phantom (Co-simulation)			0.5	$t_{co-p} n_c = 29.7$	131.1