Real-Time Full-Wave Simulations of RF Coils by Fast Integral Equation Methods

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Introduction: The electromagnetic interaction between human subjects and RF coils complicates the estimation of performance and RF power deposition. Traditional full-wave simulation approaches apply generic human models that do not accurately represent individual patients. Since the interaction is strongly patient-dependent, real-time simulations that are based on patient-specific models are desired. This study presents a full-wave real-time simulation methods based on fast integral equation method. Evaluation on a 32-channel receive coil using a dual CPU computer suggests accurate performance with computational time of less than four minutes. A promising application is per-patient specific absorption ratio (SAR) estimation.

Methods: The foundation of the fast integral equation method is the surface equivalent theorem. This approach is highly accurate, especially in modeling curved RF coils [1]. The major computational task here is to build and invert the so-called impedance matrix. The impedance matrix is dense in general and higher simulation speed can be achieved by matrix compression.

Matrix compression starts from dividing a geometrical model into sub-domains (Fig.1). It was recognized that interactions between non-touching sub-domains are rank-deficient [2]. Thus matrix blocks representing their interactions can be succinctly compressed as the outer-product of two low-rank matrices. Compressed matrices result in much less memory cost and CPU time during matrix inversion. In typical 7.0 Tesla (298 MHz) head imaging problems, it was observed that more than 80% of the impedance matrix can be compressed.

Full-pivoted LU decomposition is rank revealing and can be used for matrix compression. We applied its variant, i.e., the column-pivoted Crout algorithm, in this study. The distinct advantage of the Crout algorithm is that a matrix can be decomposed on-the-fly. This is important in efficiently generating the impedance matrix. Sub-domains can be organized further in a multi-level fashion. This results in more computational savings. Instead of organizing sub-domains by geometrical proximity, we implemented an approach based on topological proximity. This is advantageous because the size of each higher-level domain can be precisely controlled for modeling irregular objects.

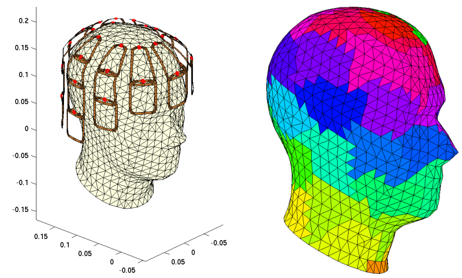

Fig. 1: The 32-channel coil model (left) and the domain-decomposed generic head model.

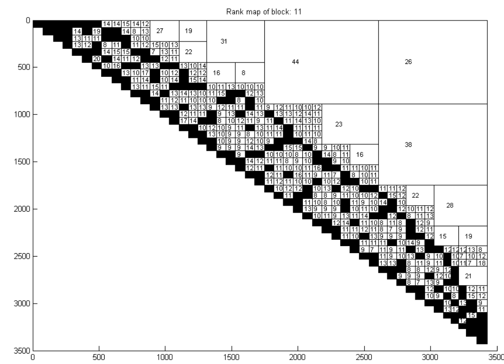

Fig. 2: Rank map of the multi-level scheme.

Results and Discussion: To demonstrate the performance, we simulated on-the-fly the geometrical noise amplification factor (g-Factor) maps of a 7.0 Tesla 32-channel receive coil (Fig. 1). The head model was obtained by a linear transformation of the generic SAM head model to a subject's MRI scan. The head model was filled with a dielectric that has $\varepsilon_r = 43$ and σ=0.41 S/m, which resembles the white matter at 7.0 Tesla. To obtain the g-Factor map, 32 individual simulations were required to calculate the B_1 profiles of each coil element. The rank map of the compressed matrix is illustrated in Fig. 2, where dark dots represent matrix blocks that can not be compressed due to close proximity. On a 2.6 GHz AMD Opteron 256 processor, an individual simulation required about 30 seconds and 120 MB RAM. The 32 simulations were organized in such way that the impedance matrix related to the head model was not recompressed for different channels. With dual processors, the 32 simulations finished within four minutes. Since the RAM requirement is low, we expect even faster simulation speed with more CPUs available.

Experiments were performed on a General Electric (Waukesha, WI, USA) 7.0 T whole body scanner using a 32-channel receive coil array designed in cooperation with, and built by, Nova Medical Inc. (Wilmington, MA, USA). Axial multi-slice data were acquired using a gradient-echo sequence with 128 x 96 resolution and 256 x 192 mm x field-of-view (FOV) using a repetition time of 3.0 s and an echo time of 9.6 ms. Both thickness and spacing of the 128 slices were 1 mm. Image reconstruction and computation of g-Factor maps (for rate 2, 3 and 4 in the left-right direction) were performed as described earlier [3]. The brain mask used to compute the coil sensitivity maps, and subsequently the g-Factor maps, was extrapolated over 4 mm. Furthermore, the acquired data were truncated in order to reduce the FOV to 230 x 156 mm^2 so that it narrowly encompassed the head. The simulated and measured g-Factor maps at acceleration rate four are shown in Fig. 3. The same gray scale range of 1.0-3.0 was used for both images. As can be seen, they match each other closely. A comparison of the simulated and measured average g-Factors for this slice at different acceleration rates are listed in Table III, where a maximum difference of about 5.5% is observed. A head model that resembles the subject's head more closely would presumably improve the correlation with experimental data. More realistic model can be obtained by non-linear transformation and this is currently under investigation.

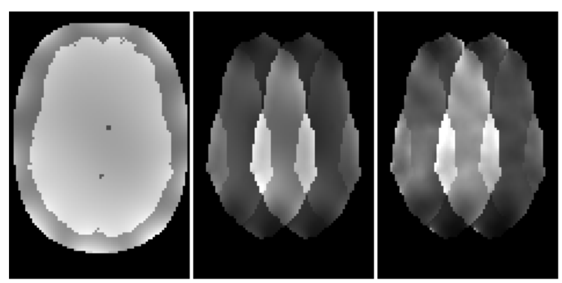

Fig 3: The masking of simulation results with the brain profile (left), the simulated (middle) and the measured (right) g-Factor maps at acceleration rate four.

Acceleration Rate	2	3	4
Simulated	1.01	1.14	1.73
Measured	1.02	1.20	1.83

Table 1: Simulated and measured average g-Factors at different acceleration rates.

Conclusion: We presented a fast integral equation method for real-time patient-dependent simulations of high-field RF coils. Results demonstrated its efficiency and accuracy. It can be applied to other patient-dependent full-wave simulations, such as SAR estimations, as well.

References: 1) Phys. Med. Biol. 51:3211-3229(2006). 2) J. Comp. Phys. 60:187-207(1985). 3) Magn. Reson. Med., 48:1011-1020(2002).