Implementation of high frequency MRI coil arrays entirely based on design-by-simulation

L. Del Tin[1], A. Peter[1], and J. G. Korvink[1]

[1]Department for Microsystems Engineering-IMTEK, University of Freiburg, Freiburg, Germany

Introduction

The use of high frequency scanners and coil arrays for MRI signal reception lead to a higher image resolution and signal-to-noise ratio [1, 2]. Such RF receiver systems are typically composed of a complex 3 dimensional (3D) distribution of many coils, operated at high frequency. The conventional approach for coil implementation, based on consecutive coil design, test and modification is for such systems not trivial and inefficient. In this work, we present a novel methodology for the implementation of high frequency 3D RF coil arrays, from device conception to their use for imaging, entirely based on simulation. The proposed procedure is applied to the realization of a cylindrical 4-coil array and its accuracy is evaluated by comparison with measurement results.

Methods

The simulation of 3D coils can be carried out using a combination of 3D finite element (FE) simulation and planar method of moments simulation. The inductive parts of the system, responsible for signal reception, are simulated using the 3D full wave FE based solver Ansoft HFSS™. For the additional circuitry needed for matching, decoupling and detuning of the coils, the 2.5D full wave solver Agilent Momentum© is used, which implements the method of moments (MoM). The S-parameters models extracted from the two simulations are linked in a circuit simulator, where the values of the lumped components, needed for a correct electrical behaviour of the coils, are calculated.

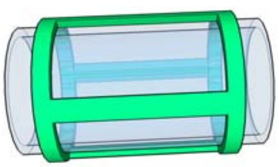


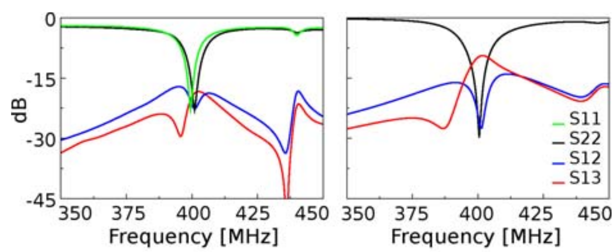
Fig. 1 Simplified model of a cylindrical 4-coil array.

FE simulations allow to monitor with good accuracy the sensitivities of the coils, eventually in presence of a sample, and to optimize their shape and dimension on the basis of field maps. The MoM simulation computes very fast the properties of any however complex flat configuration of conductors. The combination of the two methods enables the simulation of the complete coil accurately and reasonably fast. Figure 1 shows a simplified model of the coil array for 9.4T, which has been used for testing the approach. Four coils (green area in figure 1) are arranged side by side around a hallow cylinder, sharing the neighboring conductor. A volume with the material properties of silicon oil is introduced to describe the sample.

Results

The conductive tracks of the designed 4-coil array were realized on a flexible substrate and wrapped around a PEEK hollow cylinder of the proper diameter. The array was then complemented with commercial non-magnetic components, with values as established from the simulation. Fixed capacitors were used for decoupling. Two control voltages have been used to tune and match all the coils of the array.

The array was electrically tested with a network analyzer. In figure 2, measured and simulated S-parameters are reported. From the measured data, it can be seen that the coils can be tuned and matched at 400MHz, with a small frequency shift between the resonance peaks. Each coil of the array was then used as receiver coil in an imaging experiment inside a Brucker BioSpec 94-21 Magnet, with a silicone oil bottle as a sample and a Brucker linear volume coil as transmit coil. Figure 3 reports the measured signal intensities of a single coil for an axial and a transversal slice (figure 3(b) and 3(d), respectively), which are in good agreement with the computed sensitivities of the coil (figure 3(a) and 3(c)).



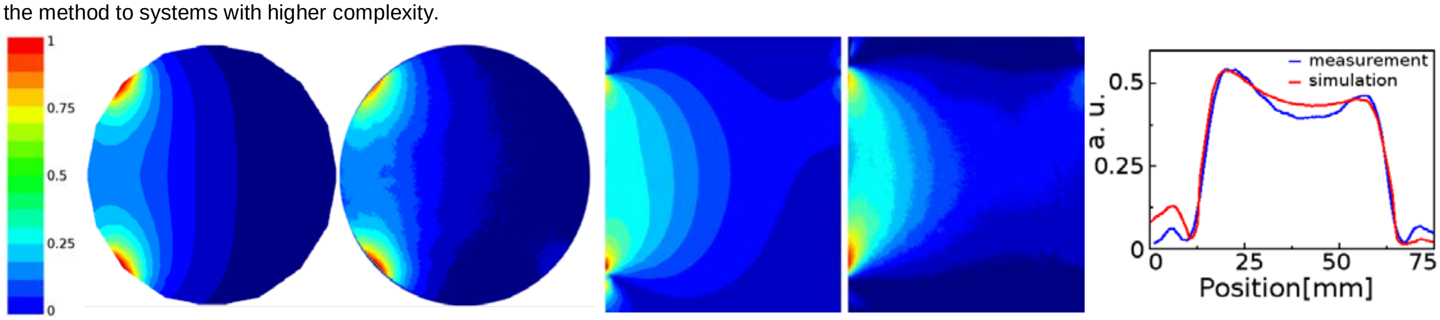
(a) *(b)*

Fig. 2 Measured (a) and simulated (b) S-parameters of the 4-coil array. Coil 1 and 2 are adjacent, coil 3 is opposite to coil 1.

Discussion

The simulation results showed good agreement with the performed measurements. The small frequency shift between the resonance peaks of the coils is due to the tollerances on the values of the capacitors. Simulated characteristics present sharper resonance peaks and a higher decoupling between adjacent coils, as well as a lower decoupling between opposite coils. The qualitative profile of the S-parameters is however captured by the simulation. Intensity maps are correctly predicted by the simulation. Wave-propagation effects are also well described by the simulation, as it is demonstrated from the asymmetry in the transversal sensitivity profile of the coils (see figure 3(e)).

The application of the proposed simulation approach enabled the straightaway implementation of a coil-array in a reasonable time and without recurring to any empirical adjustment of the array parameters. The computational cost of the FEM simulation represents the only constraint on the application of the method to systems with higher complexity.



(a) *(b)* *(c)* *(d)* *(e)*

Fig. 3 Simulated sensitivity of a single coil in the axial (a) and transversal (c) planes, compared with the achieved axial (b) and transversal (d) signal intensity. Sensitivity profile and intensity profile along a line in the transversal slice, at 5mm distance from the sample surface, are plotted in (e). Simulation and measurement data are reported to the same scale and normalized (arbitrary units).

Acknowledgements: This work is a part of the INUMAC project supported by the German Federal Ministry of Education and Research, grant 13N9208.

References: [1] Roemer et ali, Magnetic Resonance in Medicine 16, 192-225 (1990); [2] Hoult at ali, J. Magnetic Resonance 24, 71-85 (1976).