Impact of Matching Capacitors in SAR Evaluation for a 7T Endo-Rectal Coil

Jinfeng Tian[1], Lance Delabarre[1], Greg Metzger[1], and J. Thomas Vaughan[1]
[1]*U. of Minnesota, Minneapolis, Minnesota, United States*

TARGET AUDIENCE: Scientists and engineers interested in MRI RF safety, especially people who needs accurate SAR evaluation with numerical tools.

PURPOSE: The matching network is a significant component of MRI RF Coils. By properly transforming the impedance of the coaxial cable to the impedance of the RF coil, it maximizes power transfer to RF coils, eliminating the need for substantially high power RF amplifiers, and greatly reduces the possibility of circuit component breakdown by minimizing standing waves and spatially local high voltages and currents associated with the standing waves.

The importance of matching network is not necessarily reflected in RF coil simulations. When the matching network is located outside the main resonance path of the RF circuit, which is true for most RF coils for MRI use, inclusion of matching network in RF simulation can be unbearably cumbersome and significantly raise the computational effort. Consequently, the matching network is removed from some RF simulations to achieve results within a reasonable time frame. In these cases, the input RF power is normalized to emulate a "perfect match" for SAR estimation.

There are reasons to include the matching network in simulations. Although the ideal electrical components used in simulations are not at any risk of being damaged, an assessment of the local currents from simulation can predict the voltages, currents or power that individual physical capacitors will experience. From this, properly rated electrical components can be implemented to avoid component failure. The matching network may have an impact on the local SAR values, especially when the simulated coils are designed for use inside or very close to MR subject/patient body.

This abstract presents a case study of the impact of the matching network on the SAR estimation for a 7T endo-rectal coil.

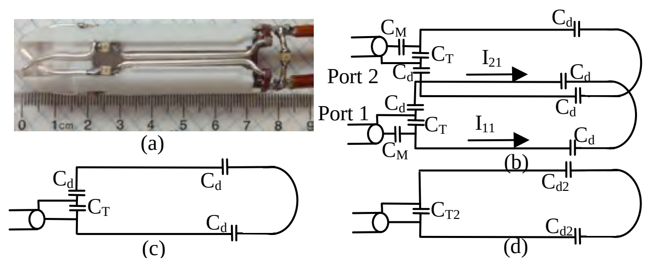

Figure 1. The endo-rectal prostate coil (a), with the original circuit (b), and simplified circuits without matching capacitor (c,d), was simulated for comparison. In circuit presented in (b) and (c), $C_T \approx C_d$ for balanced drive, and in (d) $C_{T2} \approx C_{d2}$ to generate uniform RF field. I_{21} is the induced current on loop 2 when loop is driven at port 1. Although only one loop is present in c and d, both loops were included in simulation.

METHODS and MATERIALS: The endo-rectal coil (Fig. 1a) for 7T MRI prostate imaging consists of two 1.5 cm x 7cm loops[1]. Each loop has 4 capacitors to uniformly distribute RF fields and one matching capacitor to minimize RF reflection. RF circuits, with and without matching capacitor, were also simulated (Fig. 1b-c).

The RF simulations were carried out with Finite Difference Time Domain method (XFdtd, Remcom, PA)[2]. The loops, which are mounted on the surface of a polycarbonate cylinder (5 mm o.d.) and shielded from the tissues with a thin hollow Teflon cylinder enclosure, were simulated with isotropic resolution of 2.5 mm. For SAR estimations, Duke, a 34-year adult male model (180 cm, 72 kg) from the Virtual Family, was imported with same resolution as the RF coil[3]. The tissue properties were adjusted to 296.5 MHz and the body tissues around the coil enclosure were locally modified to achieve high accuracy.

Both loop coils were tuned individually. The two loop coils were decoupled such that S21=-16dB. RF shimming was performed for all three circuits separately.

All results were normalized such that a total of 1 watt RF power was dissipated within the body model. Coil decoupling was assessed in two ways: S21, and current ratio as $20 \cdot \log10(I_{21}/I_{11})$, where I_{21} is the induced current on Loop 2 when Loop 1 is active and has a current of I_{11}.

Results and Discussions: $|\mathbf{B}_1^+|$ and SAR values are listed in Table 1, and their distributions on trans-axial and sagittal slices across prostate are presented in Figure 2. 1). $|\mathbf{B}_1^+|$ within the prostate is much stronger with RF shimming for all the cases, showing the effectiveness of the RF shimming. 2). Compared with the original circuit, both the simplified circuits in Fig. 1c and Fig. 1d have smaller global maximum 1g and 10g SAR, and smaller maximum 1g and 10g SAR at the transverse slice across the matching capacitors. While the global maximum SAR is mainly affected by all RF circuit components, the local SAR variation is largely due to the impact of the matching capacitors. 3). The discrepancy between the two decoupling assessment methods deserves attention when S21>-17dB.

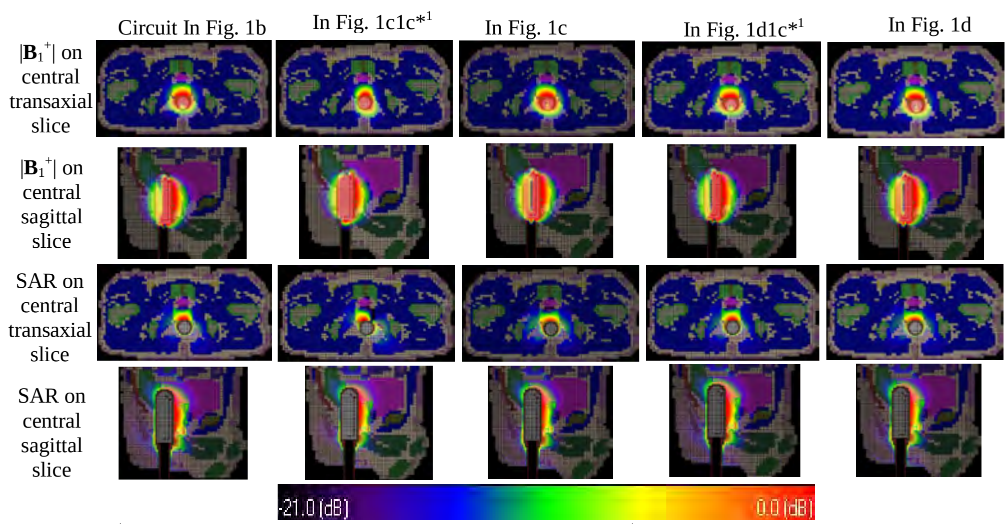

Figure 2. $|\mathbf{B}_1^+|$ and SAR distributions in central slices. 0dB is 5µT for $|\mathbf{B}_1^+|$, and 10 Watt/kg for the un-averaged SAR.

Conclusion: Matching network is not negligible in numerically evaluating SAR for RF coils inside or very close to patient/subject body.

References: 1. Metzger G, MRM 64: 1625-39. 2. Collins CM, MRM 40: 847-56. 3. Christ A, Phys. Med. Biol 55: 1767-1783. **Acknowledgement:** NIH-NIBIB 2R01 EB007327, NIH-NIBIB 2R01 EB006835, P41 EB015894, S10 PR029672, Keck Foundation

Table 1. SAR Statistics around the Coil and Prostate

Circuit	$\lvert\mathbf{B}_1^+\rvert$ at Prostate Center (µT)	Max 1g/10g SAR (W/kg)*[2]	Max 1g/10g SAR on the Matching Slice (Trans-axial) (W/kg)	Decoupling Current Ratio (dB)	Decoupling S21 (dB)
Fig. 1b	3.42	8.94/4.94	5.10/3.24	-12.19	-16.21
Fig. 1c*[1]	3.06	9.14/4.56	4.35/2.65	-22.68	-23.78
Fig. 1c	3.21	6.61/3.70	4.62/2.43	-22.68	-23.78
Fig. 1d*[1]	3.31	8.12/4.62	4.43/2.89	-21.83	-24.04
Fig. 1d	3.24	7.40/4.22	4.05/2.65	-21.83	-24.04

***[1]**: Coil was driven with the same phase set as *[1] for the circuit in Fig 1b. ***[2]**: Global maximum SAR occurs on the trans-axial slice at the level of C_d.