Calculations of RF Power Absorbed by the Portions of a Human Body Inside and Outside a Body Coil's Imaging Volume at 64 and 128 MHz

W. Liu[1], C. M. Collins[1], Q. X. Yang[1], M. B. Smith[1]

[1]Radiology, Penn State College of Medicine, Hershey, PA, United States

INTRODUCTION: The square root of RF power absorbed in the sample in MRI is proportional to the received noise assuming sample noise dominates (1). In human MRI, part of the RF power is necessarily absorbed by the portion of the body inside the coil to maintain the B1 field needed for imaging. The portion of the body outside the coil absorbs another part of the RF power. Knowledge of the distribution of the RF field and the absorbed power in these two portions may facilitate design of RF coils that limit unnecessary power absorption resulting in improved SNR.

METHODS: A shielded highpass body size (63 cm coil diameter, 68 cm shield diameter, 70 cm length) birdcage coil loaded with a human body was modeled at 64 and 128 MHz with the Finite Difference Time Domain (FDTD) method. The inner bore of the stainless steel magnet casing (90 cm diameter, 242 cm long) was also modeled. A Perfect Electric Conductor (PEC) boundary was used to simulate the RF screen (2). The coil was put at the center of the magnet and the center of the coil was 10 cm above the patient bed. The central axial plane of the coil crosses the center of the subject's heart. The spatial resolution was 5 mm in all three dimensions. The coil was tuned to the desired frequencies with lumped-element capacitors and was driven in quadrature. The steady-state RF electrical field (E) and magnetic field (B_1) were calculated and the power absorbed in the body was calculated using the formula $P_{abs}=\sigma E^2$ where P_{abs} is the absorbed power, σ is the conductivity of the tissue, and E is the magnitude of the electrical field (3). The coil's imaging volume was defined as the space between the top and the bottom axial planes of the coil, whose locations are indicated by the dashed lines in Figure 1. The power absorbed by the portion of the body inside the coil's imaging volume, P_{absin}, and by the portion of the body outside the coil's imaging volume, P_{absout} were calculated. The B_1 magnitude distribution on the central coronal plane was also calculated.

RESULTS: The magnitude distribution of the B_1 field on the central coronal plane is given in Figure 1. The percentage of the total power dissipated in the body portion inside and outside the coil's imaging volume is given in Table 1 at both frequencies.

DISCUSSION: At 64 MHz, 95.76% of the input power is absorbed by the portion of the body inside the coil's imaging volume and 4.24% by the portion outside. At 128 MHz, the percentage of the input power absorbed by the portion outside the coil's imaging volume is about twice as much as that at 64 MHz but still remains below 10% of the input power. Stronger B_1 field magnitude is seen in the subject's lower body outside the coil at 128 MHz than at 64 MHz on the central coronal plane as shown in Figure 1. This stronger field distribution outside of the coil at the higher frequency might be related to more power radiated out of the coil (2,4), and eventually absorbed by the portion of the body outside of the coil.

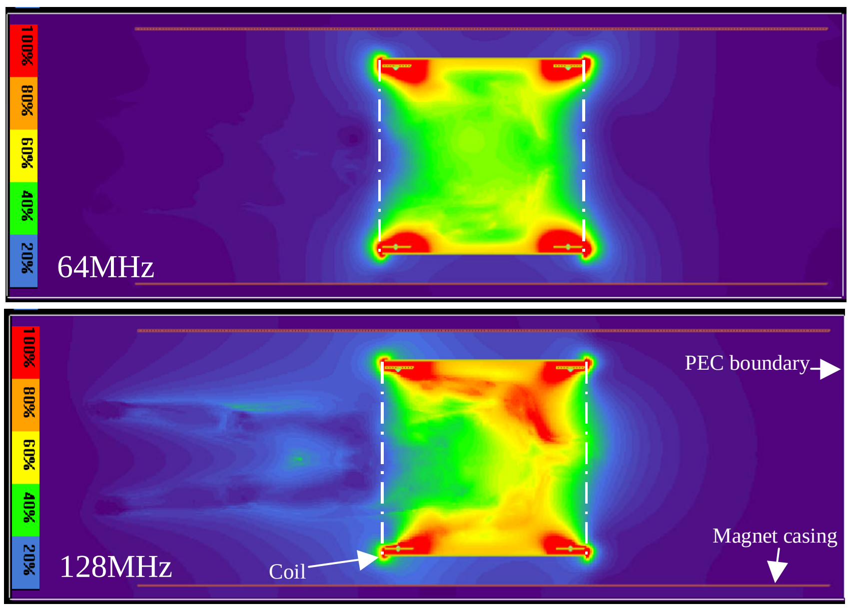

Figure 1. $|B_1|$ distribution on the central coronal plane for 64 and 128 MHz. The space between the two dashed lines is the imaging volume of the coil. At each frequency, the scale maximum is twice the field strength at the coil center.

Table 1. Percentage of input power absorbed by the body portion inside the coil's imaging volume, $P_{absin}/(P_{absin} + P_{absout})$ and by the body portion outside the coil's imaging volume, $P_{absout}/(P_{absin} + P_{absout})$

Frequency	$P_{absin}/(P_{absin} + P_{absout})$	$P_{absout}/(P_{absin} + P_{absout})$
64 MHz	95.76%	4.24%
128 MHz	91.25%	8.75%

REFERENCES:
1) WA Edelstein et al. MRM 3:604-618(1986)
2) CM Collins et al. Proc. ISMRM p714 (2003)
3) CM Collins et al. MRM 45:684-691 (2001)
4) W Liu et al. Proc. ISMRM p2393 (2003)

ACKNOWLEDGEMENT: We would like to acknowledge Dr. David Doty and Dr. Cecil Hayes for discussions leading to this work. Funding for this work was provided through NIH EB000454.