Evaluation of Displacement Currents and Conduction Currents in a Close Fitting Head Array with High Permittivity Material

ChristopherM. M. Collins[1,2], Giuseppe Carluccio[1,2], Manushka Vaidya[1,2], Gillian Haemer[1,2], Riccardo Lattanzi[1,2], Graham C. Wiggins[1,2], Daniel K. Sodickson[1,2], and Qing X. Yang[3]

[1]*Center for Advanced Imaging Innovation and Research (CAI2R), New York University School of Medicine, New York, NY, United States,* [2]*Bernard and Irene Schwartz Center for Biomedical Imaging, New York University School of Medicine, New York, United States,* [3]*Center for NMR Research, Penn State College of Medicine, Hershey, PA, United States*

Audience: All interested in improved performance and safety of MRI through engineering of RF Coils and Fields.
Purpose: Explain mechanisms by which high-permittivity materials (HPMs) can improve performance of close-fitting coils and arrays.

Introduction: In recent years, HPMs (with ε_r>>1) have shown great promise for improving SNR and reducing SAR in a number of applications at 3T and 7T (1, 2). While most work with HPMs has been focused on improving SNR or transmit efficiency for a relatively small region within a much larger coil or array, recent work has also shown that HPMs can improve performance of smaller coils very near the subject (), as well as arrays of such coils for the entire region of the anatomy they encompass (3). Here we discuss mechanisms for this with simulated distributions of magnetic and electric fields as well as conduction and displacement currents.

Theory: According to Maxwell's 4th Equation, $\nabla \times \mathbf{B} = \mu(\sigma\mathbf{E} + j\omega\varepsilon\mathbf{E}) = \mu(\mathbf{J} + \mathbf{D})$, such that both conduction currents ($\mathbf{J}=\sigma\mathbf{E}$) and displacement currents ($j\omega\mathbf{D}=j\omega\varepsilon\mathbf{E}$) through space can contribute to the magnetic field. In MRI today, coils are typically designed with consideration only of conduction currents in discrete copper elements. These coils are associated with sizeable electric fields throughout space, which can produce significant displacement currents as well by using HPMs.

For a conductive segment of an RF coil near an HPM, it is possible (using inductance of the coil segment L, required voltage to drive a current I at frequency ω, and resulting near-field E and B) to calculate the displacement currents in the HPM and show that they will add to the magnetic fields within the nearby sample for a given coil current. Additionally, in the near field of the conductive segment (having a different E/H ratio than the far field) the optimal materials for impedance matching can have a higher permittivity than those of human tissues (4). These effects lead to stronger fields in the region of the sample encompassed by the HPM. Also, because the displacement currents are more distributed than the conductive currents, the result is safer (inducing lower peak local SAR) than if the conductors were simply moved closer to the subject. The net result is stronger coupling to the region of interest (ROI) relative to the rest of the body, thus higher SNR and lower SAR.

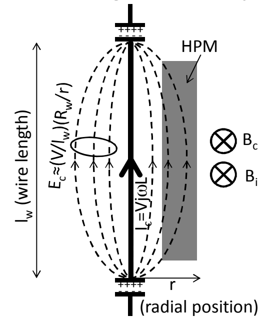

Figure 1. Schematic illustration of a conductive segment near a high permittivity material (HPM), the current in the coil I_c, conservative electric fields through space E_c, and magnetic fields produced by I_c (B_c) and induced by E_c (B_i).

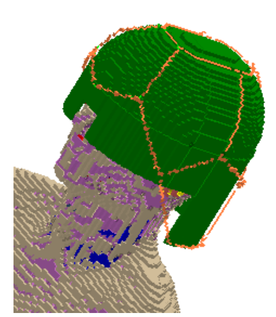

Figure 2. Geometry of transmit/receive array model based roughly on existing 7T array (7). For single-coil simulations, the coil most visible on the left side of the head is driven. Coil former (green) is alternately assigned properties of air or realistic HPM.

Methods and Results: An 8-element, 7T, transceive array was simulated on a 5mm-thick helmet-shaped former about the head of a numerical model of the human body ("Duke") within a conductive magnet bore, as shown in Figure 2. The former was alternately assigned dielectric properties of air or a slurry of Calcium titanate powder and water with ε_r of 107 and σ of 0.08S/m at 300 MHz (5).

When a single coil is driven (Fig. 3), it is seen that use of the HPM increases transmit and receive efficiency to the ROI surrounded by the HPM while reducing coupling (E and B fields) to the rest of the body. It has previously been shown that this results in an overall improvement in SNR for the cerebrum in excess of 45% (3), and that the presence of the HPM does not reduce efficacy of RF shimming during transmission (6), but increases transmit efficiency resulting in markedly reduced max. local, head-average, and whole-body SAR levels for a desired B_1 strength (3, 6).

Plots of $|\mathbf{B}_1^+|$, $\|\mathbf{E}\|$, modulus of the conduction current $\|\mathbf{J}\|$ (where $\mathbf{J}=\sigma\mathbf{E}$), and modulus of the displacement current $\|\omega\mathbf{D}\|$ (where $\mathbf{D}=\varepsilon\mathbf{E}$) for cases with and without the HPM former are shown in Fig. 4. Here RF shimming was performed to produce constructive interference in $|\mathbf{B}_1^+|$ at the center of brain and fields were normalized to produce 1μT average $|\mathbf{B}_1^+|$ in brain on the plane shown. In the Table, the maximum values of $\|\mathbf{J}\|$ in the copper conductors and for $\|\omega\mathbf{D}\|$ in the HPM, as well as the sum of all $\|\mathbf{J}\|$ in the copper conductors and of all $\|\omega\mathbf{D}\|$ in the HPM are given for both simulated cases with the same normalization.

With the HPM former present, the maximum conduction current in space is about 4 times larger than the maximum displacement current, but an integration of displacement current through the HPM former gives a value 25% larger than the integration of conduction current in the copper conductors, suggesting a greater contribution to $|\mathbf{B}_1^+|$ in the ROI from the displacement currents than from the conduction currents.

It is clear that when the HPM former is present, the distributed displacement current $\|\omega\mathbf{D}\|$ in the former results in a need for lower conduction currents $\|\mathbf{J}\|$ in the copper coils, as well as lower electric fields $\|\mathbf{E}\|$ throughout space. Additionally, the distributed displacement current $\|\omega\mathbf{D}\|$ in the HPM former results in a more homogenous $|\mathbf{B}_1^+|$ distribution than the discrete copper coils can produce.

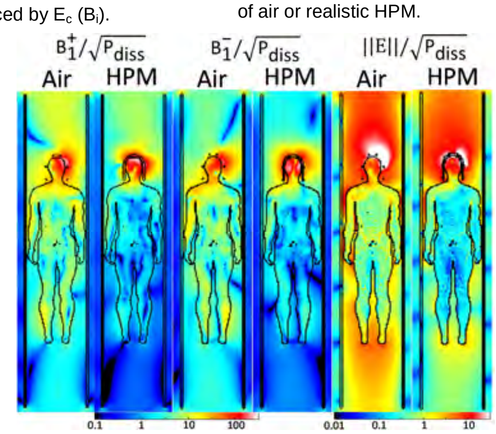

Figure 3. Transmit efficiency (nT/√W), receive efficiency (nT/√W), and normalized E fields (V/m/√W) throughout the subject and bore for a single surface coil adjacent the head with and without the presence of an HPM coil former (green in Fig. 3). All fields are stronger in the ROI (head region) and lower elsewhere in the body, indicating better sensitivity to signal in the ROI and less sensitivity to noise from the rest of the body (better SNR), as well as less total subject heating during transmission, when the HPM former is used.

Discussion: Strategic use of high-permittivity materials in MRI can improve SNR and reduce SAR of conductive coils and arrays (1-6). The mechanisms involved are in addition to those of field strength and multi-channel transmission or reception. The results shown here coupled with previous ones (1-7) show that HPMs can improve MRI significantly, and should be developed further.

The HPM layer simulated here is 5mm thick. In any commercial coil there is at lerast this much space available between the copper conductors and insulative casing, so no more space in the bore or greater distance between conductive coils and sample is required.

In addition to the mechanisms mentioned here, ongoing work indicates that the presence of a dielectric layer can bring a circular coil of reasonable size approach the ultimate intrinsic SNR (7,8) or a given array approach ultimate intrinsic SAR (9), indicating value of HPMs in arrays with a reasonable number of coils.

Significant challenges remain in producing rigid materials with the size, shape, and dielectric properties needed to create HPM formers like that simulated here, and in ensuring adequate coil performance (tuning matching, decoupling, etc.) in close proximity to the HPM. Ongoing efforts are designed to address these challenges.

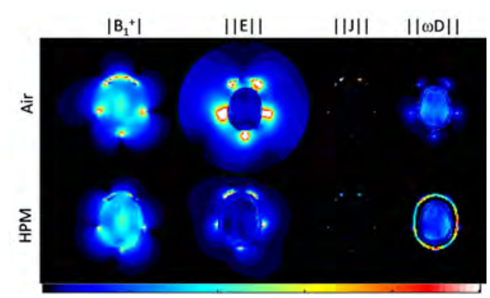

Figure 4. Fields and currents on axial plane through the center of brain. Linear scale max is 3μT for |B1+|, 1000V/m for ||E||, 5000A/m^2 for ||J||, and 250A/m^2 for $\|\omega\mathbf{D}\|$. Fields are normalized to 1μT average in brain on this plane.

Former material	Max(||J||) in coil (A/m^2)	Sum(||J||) in coil (A/m^2)	Max(||ωD||) in former (A/m^2)	Sum(||ωD ||) in former (A/m^2)
Air	13290	3.30x10^6	-	-
HPM	7271	1.20x10^6	1953	1.50x10^6

References:
1. Webb AG. Concepts in MR 2011;38A:148-84
2. Yang QX *et al*., MRM 2011;65:358-362
3. Brink WM and Webb AG. MRM 2014;71:1632
4. Collins CM *et al*., 2014 ISMRM p. 40
5. Carluccio G *et al*., 2013 ISMRM, p. 4374
6. Luo W *et al*., MRM 2013;70:269-275
7. Collins CM *et al*., 2014 ISMRM, p. 1340
8. Lattanzi R *et al*., 2014 ISMRM p. 4818
9. Vaidya M *et al*., 2015 ISMRM, submitted
10. Haemer G *et al*., 2015 ISMRM, submitted