Patient-specific SAR estimation for the 8-channel radiative antenna array at 7 T prostate MRI

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Introduction: The specific absorption rate (SAR) is an important safety concern in ultra-high field MRI (>7T) (1). More particularly, local SAR rather than global SAR, is the limiting factor of an external surface coil array at 7 T (2-3). Local SAR cannot be measured and therefore, safety limits must be determined by electromagnetic simulations on patient models. The question arises whether an accurate model of the scanned subject is required, or whether one generic patient model would be sufficient. In this study, we will answer this question for our newly developed transceive surface array for prostate imaging at 7 Tesla (4). The array consists of so-called radiative antennas, which are effectively dipole antennas on a substrate of high-dielectric ceramic. For this array, we determined the patient-specific worst-case local SAR in 3 different patient models, by calculation of the largest eigenvalue of the Q-matrix for each voxel. The models that we used vary both in size and fat/muscle ratio. In this way, we will obtain an impression whether a generic model is sufficient for worst-case local SAR estimation of an external surface array.

Methods: The CT images of 3 prostate patients (born in 1938-1943) were segmented by using iSeg (IT'IS, Zurich, Switzerland). The 3D segmented abdomen images consist of bone, muscle, fat and skin (Figure 1). The patient abdomen circumferences ranged from 25×38 cm^2 to 33×44 cm^2. After exporting the segmented patient models to SEMCAD X (SPEAG, Schmid & Partner Engineering, Zurich, Switzerland), FDTD simulations with the 8-element radiative antenna array were performed. To ensure the direct contact of the antenna elements to the patient body, additional skin (4 mm) and and fat (6 mm) layers were added under the upper elements. The grid resolution was non-uniform (with minimum voxel dimensions of 2.4 × 0.3 × 0.3 mm^3). The substrate of each radiative element has dimensions 6 × 5 ×15 cm (width × height × depth) with a relative electrical permittivity of 37, and the conductors of the dipole antenna is 1 cm wide and 6 cm long. The elements were tuned to 298.2 MHz by using an inductor of 5 nH. Impedance matching was ensured by chosing the source impedance equal to the (real) tuned impedance of the element. The E-field distribution for each single array element was determined by 8 individual simulations. After exporting the voxelized model and E-fields of these simulations to Matlab (The MathWorks Inc., Natick, MA), they were cropped and interpolated at 2 mm resolution and exported to hdf format. The eigenvalues and eigenvectors of the 10-gram averaged SAR values were calculated for all patient models within in-house developed tools written in C++.

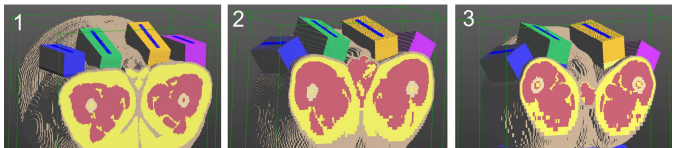

Figure 1: 8 element radiative antenna array placed on 3 segmented patient models.

Results and Discussion: Figure 2 (a,b,c) shows the eigenvalue maps of each model. These maps represent the maximum 10 gram averaged SAR that can be deposited on each voxel by 1 W total power, given the worst possible shim settings. For all models, the maximum SAR hotspots are observed in the skin just under the elements. These results show that for the same sequence parameters, model 1 will be exposed to a maximum SAR that is 2.5 times as high as for model 3. This has severe consequences if model 2 is meant to act as a generic model for the entire patient population: Based on model 2, the local SAR limit of 20 W/kg (5) is not exceeded if the duty cycle remains below 0.006 (e.g. pulse length of 5 ms, TR of 0.88 s). However, if one applies this duty cycle in the scan of model 3, the estimated SAR reaches 49 W/kg which severely violates the guidelines (Table1). Therefore, we conclude that the SAR analysis for the surface array can not be based upon just one patient model. This is likely also the case for other surface array designs. Further analysis, including more patient models, will show whether we can come up with a set of generic models. The SAR exposure of a new patient will then be calculated by the model that comes closest to the patient anatomy of the given patient. Furthermore, note that the eigenvector that belongs to the maximum eigenvalue has one dominant index that corresponds to the closest array element, indicating that the maximum local SAR is mainly governed by the power of that element (Figure 2 d,e,f). However, other elements can contribute significantly (between 17 and 30 %). A local SAR monitoring concept based on power delivery per channel is thus feasible but requires a safety margin of at least 30 %.

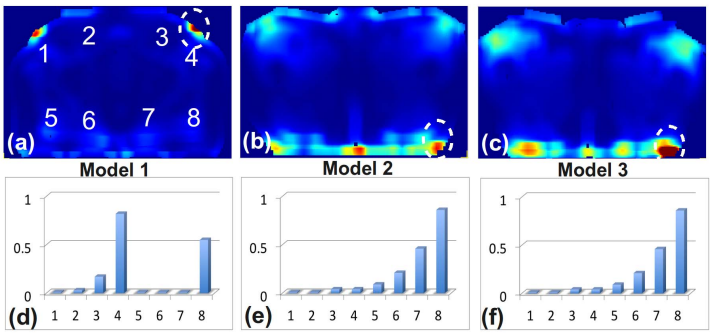

Figure 2: (a,b,c) The SAR eigenvalue maps of 3 different patient models where the maximum eigenvalue is shown in circle. (d,e,f) eigenvectors that belong to the maximum eigenvalue in the image (indices correspond to array elements,)

SAR based on model	resulting max. SAR exposure		
	1	2	3
1	20 W/kg	14.2 W/kg	34.4 W/kg
2	28 W/kg	20 W/kg	49 W/kg
3	12 W/kg	8.4 W/kg	20 W/kg

Table 1: Worst-case SAR estimation per model if one model is chosen as a generic model for the others.

Conclusion: A generic model is not sufficient for local SAR calculations of surface arrays for MR imaging at 7 Tesla. This study will be extended by including more patient models.

References: (1) Collins C M *et al* 2001 *Magn. Reson. Med.* **45** 692 (2) Metzger G J *et al* 2010 *Magn. Reson. Med.* **64** 1625 (3) Bitz A K t al *Proc. Intl. Soc. Mag. Reson.* **19** 490 (4) Raaijmakers A J E *et al* 2011 *Magn. Reson. Med.* **66** 1488 (5) Center of Devices and Radiological Health. Criteria for significant risk investigations of magnetic resonance diagnostic devices 2003.