

# Patient-specific SAR models and in vivo validation

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## Introduction

For parallel transmission, traditional safety concepts need to be re-assessed, as the local specific absorption rate (SAR) becomes a limiting safety constraint [1] that depends on the pulse design as well as on the individual patient anatomy. Whereas the whole-body SAR can be monitored during the scan, the local SAR cannot be measured in clinical practice. Instead, numerical simulations are usually performed for safety approval. In recent years, a number of generic body models have been developed for this purpose (e.g. [2,3]). However, the local SAR is dependent on the subject-specific anatomy and the body pose inside the scanner. Furthermore, these models have, to our knowledge, never been validated *in vivo*. Therefore, it is highly desirable to provide additional, subject-specific body models for comparative SAR studies. In this work, a method for generating dielectric body models from MR scans was developed, consisting of the following steps: (1) Whole-body images are acquired, (2) a water-fat separation is performed, (3) segmentation into three tissue classes is carried out, and (4) FDTD models are created. The simulation results are validated against measured  $B_1^+$  field maps.

## Methods

In a preliminary study [4], we showed that the local SAR pattern in a 3T body coil is highly robust against tissue variations, as long as the body-fat distribution is maintained. Eddy currents induced in the patient by the RF transmit field are pinched by fatty, low-conductive tissues. On the other hand, muscles and inner organs exhibit a similarly high conductivity and combining them to a single 'muscle-like' group did not result in significant errors [4]. Therefore, a fat-water separated body model will presumably give a realistic representation of the current pathways and hence local SAR in the patient.

*In vivo* scans were performed on five healthy volunteers in different body poses on a 3T MR scanner (Philips Healthcare, Best, The Netherlands), equipped with an 8 Tx-channel body coil [5]. Whole body (head-to-toe) images were recorded in a multi-station scheme (3D multi-echo FFE, TR/TE<sub>1</sub>: 5.6/2.3 ms,  $\alpha = 10^\circ$ ). At each station, 3 echoes at  $\Delta TE = 0.78$  ms were sampled. The pixel bandwidth was about 10 times the water-fat shift, so that the lateral shift was negligible. A spatial resolution of 5 mm in all three dimensions was used, which has been proven to be reasonably accurate for FDTD calculations of the 10g averaged local SAR for a 3T body coil [4].

Water-fat separation was performed via a 3-point Dixon reconstruction [6] for each station. The stacks were then combined to whole-body water and fat images. To obtain a tissue segmentation, the 2D histogram of the water and fat image pairs was calculated. Then, an expectation-maximization (EM) algorithm [7] was applied to fit a mixture-model. The fat and water regions were each modeled by a Gaussian distribution and an exponential distribution was used for the background. The lung was then identified using a simple region-growing algorithm inside the human body. Bones caused holes in the segmentation which were labeled as fat to represent a low conductivity. Dielectric properties of muscle, fat, and lung tissue at 128 MHz [8] were assigned to the three segments.

The final body models and the RF body coil were simulated using the finite-differences time-domain FDTD (xFDTD, Remcom Inc.) technique, highly parallelized on 2 GPUs. Simulated  $B_1^+$  fields were calculated for the quadrature drive mode.

For validation purposes,  $B_1^+$  field maps were acquired directly after the whole-body scan in a central abdominal location. A fast steady-state FFE  $B_1^+$  mapping technique (actual flip angle: AFI) [9] was applied to allow for measurements within a single breathhold (5x5 mm, 15 mm thickness, TR<sub>1</sub>/TR<sub>2</sub>: 20/100 ms,  $\alpha = 50^\circ$ ) with RF multi-channel excitation in quadrature mode.

## Results and Discussion

Whole-body imaging took about 5 minutes and was mostly limited by repeated acceleration and deceleration of the patient table. The 2D histogram-based EM segmentation converged within 30 to 45 seconds and showed to be highly robust against variations of the initial distribution parameters, such that our approach to SAR model generation seems suitable for future automation. A cross-section through a typical body model is presented in Fig. 1. Visual comparison of the models and the original images showed that the water and fat segments agree with the physiology.

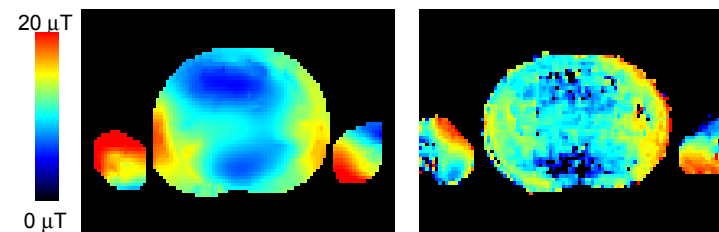
FDTD simulations of the whole-body models took about 40 minutes for each RF channel. The comparison of the FDTD result and the AFI map in Fig. 2 supports the qualitative and quantitative agreement of the wave-propagation inside the body. Both images show a higher field strength at the sides of the body and a local maximum in the center. The accuracy in the arms is reduced due to main field and gradient inhomogeneities in the outer regions. The correlation coefficients between simulations and measurements over the torso ranged up to 71%.

## Conclusion

The presented method provides a fast and straightforward way to generate subject-specific SAR models. The method allows creating a greater model diversity and facilitates *in vivo* validation of simulation results. The approach facilitates reliable subject-individualized SAR estimation which might potentially be used for real-time SAR calculations in parallel transmit MR systems. Alternatively, inter-subject comparison of simulated SAR values will provide further confidence for safety assessment.



**Fig. 1:** Dielectric body model from 3D MRI volunteer data with three tissue classes:  
Muscle : dark gray  
Fat tissue: light gray  
Lungs : white  
Resolution is 5mm, isotropic.



**Fig. 2:** The FDTD simulated  $B_1^+$  field (left) is qualitatively and quantitatively consistent with the measured AFI field map (right). Both images were calibrated to a mean value of 10  $\mu T$ .

## References

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