# Body Models of Big People for MRI Safety Assessment

Ingmar Graesslin[1], Hanno Homann[1], Holger Eggers[1], Kay Nehrke[1], Paul Harvey[2], Cecilia Possanzini[2], and Peter Börnert[1]
*[1]Philips Research Laboratories, Hamburg, Germany, [2]Philips Healthcare, Best, Netherlands*

**Target audience:** RF engineers and MR physicists

**Purpose:** The specific absorption rate (SAR) is often a limiting factor for MR imaging at higher field strength. Global and local SAR are predicted before an MRI scan from electromagnetic simulations using discretized body models to ensure the scan is SAR safe and in the limits.[1] However, the type of transmit coil used, the anatomy of the patient, and the position and orientation of the patient relative to the coil generate uncertainty in SAR simulations, which are more demanding for parallel transmission MR systems.

For SAR assessment, it is desirable to generate a wider variety of whole-body models from MRI data. To scan the subject, he or she can be moved station-by-station in feet–head (FH) direction through the isocenter. This approach overcomes the main field homogeneity restrictions of existing magnets and allows the virtual extension of the field of view (FOV) in the FH direction. However, for large subjects, such an acquisition is challenging, because restrictions in actual FOV persist even for the latest wide bore systems. To address this problem, a virtual extension in the lateral (left–right, LR) direction is needed. Therefore, in this work, we extended the conventional FH virtual FOV imaging with lateral virtual FOV imaging to scan large subjects. This approach can be implemented easily in open, non-cylindrical MR systems that allow lateral table movement in the FH and LR directions (see Fig.1). Here, we outline the basic concept, the results of volunteer experiments, and the use of the generated whole-body models for the safety assessment of very large patients.

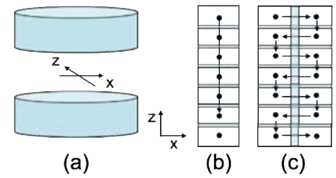


Fig.1. Scheme for lateral virtual FOV scanning. (a) Open MR systems allow table motion in two directions (z: FH, x: LR). (b) Scheme for conventional multi-station imaging, (c) added lateral virtual FOV imaging with 2 stations. Other trajectories are conceivable.

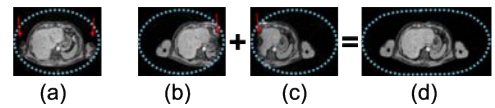


Fig.2. Principle of lateral FOV enlargement. (a) Due to the finite magnet homogeneity, the actual FOV is restricted (dotted line): see resulting image artifacts (arrows). (b,c) By acquiring multiple laterally shifted data sets, the FOV can be enlarged virtually using appropriate image fusion (d).

**Methods:** The basic $B_0$ field homogeneity problem is outlined in Fig. 2. Due to the finite FOV in LR direction, the MR signal from the arms is corrupted for a very large subject. If two or more scans, shifted in the LR direction, are performed, portions of the subject can be scanned within the homogeneity volume of the main field, so that the entire subject is covered laterally (see Fig. 2c,d). To cover the entire subject, this multi-station approach can be performed along the two spatial directions in which table movement is possible (LR, FH). Different z–x trajectories are conceivable for such an approach (e.g., the trajectories in Fig.1). The resulting data must be merged via appropriate post-processing, considering the information of the spatial overlap.

In a feasibility study, experiments were conducted on 7 healthy adults (aged 30–58 years, weighing 68–135 kg) using a 1.0T Open scanner (Panorama, Philips Healthcare, The Netherlands). 3D spoiled gradient echo (FFE) data were acquired (5 mm isotropic resolution) using a 3-point Dixon imaging sequence ($\alpha$: 10°, TR: 7.4 ms, TE: 1.7/3.7/5.7 ms) in a 2D multi-station mode (two stations in LR direction, 15 stations in FH direction, overlap LR: 125 mm, FH: 20 mm). The virtual FOV was $770_{LR}\times400_{AP}\times2000_{FH}$ mm,[3] and the total scan time was about 15 minutes. The water/fat data were separated using a flexible and robust algorithm[2] and subsequently fused to water and fat images.

The model generation was carried out as reported previously.[3] If the intensity of each voxel in the water image was greater than in the fat image, it was classified as "water;" and if the intensity was greater in the fat image than in the water image, it was classified as "fat." As a simple classifier for the lung tissue and the background an intensity threshold based on a mixture model was used. The water/fat-dominated tissues were modeled by two Gaussian distributions, and the background by an exponential distribution.[3] To obtain the three model parameters, an expectation-maximization algorithm was used. Optionally, the surface voxels of the models can be replaced with the tissue type skin.

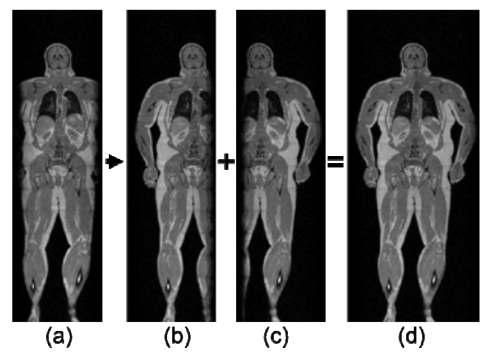


Fig. 3. 3D in-phase 2D multi-station whole-body imaging data of an XXL volunteer. Coronal reformats are shown for (a) a conventional 1D multi-station scan (note missing signal from the upper extremities). (b,c) Separate results of two 1D multi-station scans displaced in lateral direction. (d) Combined result of the 2D multi-station scan. Slight signal drop-off in the elbows indicates that three lateral stations would have been desirable in this case.

**Results and Discussion:** The MR signal of the arms of the large subject is corrupted due to the finite FOV in LR direction (Fig. 2). The in-phase images of one selected volunteer (Fig. 3) illustrate the basic feasibility of the 2D multi-station acquisition mode. The acquisition of two data sets, each appropriately shifted in the LR direction, generates a data set of the entire subject after appropriate post-processing and image fusion. Compared with conventional multi-station images (Fig. 3a), the single-run multi-station images are appropriately displaced from the magnet's isocenter (Fig. 3b,c). The fused 3D data set of the 2D multi-station scan (Fig. 3d) now covers the complete subject. The 3D water/fat resolved 2D multi-station coronal and sagittal images of one selected volunteer show a good separation of water and fat. Next to these images, the generated model is shown, which differentiates between muscle, fat, and lung tissue (Fig. 4).

SAR modeling of such subject-specific models has been shown,[3,4] and an initial analysis of SAR accuracy has been presented using simple models,[3] based on the same approach for an 8-channel 3 T body coil. However, further anatomical detail is needed for head coil simulations, and suitability for simulations at higher field strength (≥7 T) must be validated.

In addition to model generation, this approach can be used for potential basic diagnostic applications using 2D and 3D imaging techniques. Furthermore, it might be applicable to radiation treatment planning, and for PET/MR attenuation map correction.

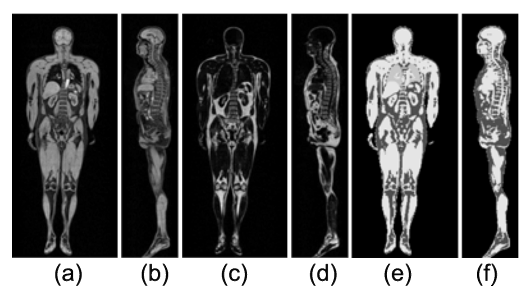


Fig.4. 3D water/fat-resolved 2D multi-station imaging results of a volunteer and corresponding body model. (a,b) Coronal and sagittal reformats of the water data. (c,d) Fat data and (e,f) model data, respectively. This model consists of fat, muscle, and lung tissue.

**Conclusion:** We present a novel approach to generate whole-body models for SAR simulations of very large subjects. The new concept uses 2D multi-station imaging with lateral virtual FOV enlargement and a water/fat-resolving imaging sequence. The main area of application is the development of patient-specific body models, which may enable improved SAR management and reduced scan time in parallel transmit MRI.

**References: 1.** International Electrotechnical Commission, *International standard, Medical equipment – IEC 60601-2-33: Particular requirements for the safety of Magnetic resonance equipment* (3rd ed.), 2010. **2.** K. M. Johnson, G. D. Leavitt, and H. W. Kayser, "Total-body mr imaging in as little as 18 seconds," *Radiology*, 1997, 202:262–267. **3.** H. Homann, P. Börnert, H. Eggers, et al., "Towards Individualized SAR Models and In Vivo Validation," *Magn Reson Med.*, 2011, 66 (6):1767–1776. **4.** I. Graesslin, H. Homann, S. Biederer, et al., "A Specific Absorption Rate Prediction Concept for Parallel Transmission MR," *Magn Reson Med.*, 2012, 68(5):1664–1674.