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Whole-body hybrid imaging concept for the integration of PET/MR into radiation therapy treatment planning

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Abstract

Modern radiation therapy (RT) treatment planning is based on multimodality imaging. With the recent availability of whole-body PET/MR hybrid imaging new opportunities arise to improve target volume delineation in RT treatment planning. This, however, requires dedicated RT equipment for reproducible patient positioning on the PET/MR system, which has to be compatible with MR and PET imaging.

A prototype flat RT table overlay, radiofrequency (RF) coil holders for head imaging, and RF body bridges for body imaging were developed and tested towards PET/MR system integration. Attenuation correction (AC) of all individual RT components was performed by generating 3D CT-based template models. A custom-built program for \( \mu \)-map generation assembles all AC templates depending on the presence and position of each RT component. All RT devices were evaluated in phantom experiments with regards to MR and PET imaging compatibility, attenuation correction, PET quantification, and position accuracy. The entire RT setup was then evaluated in a first PET/MR patient study on five patients at different body regions.

All tested devices are PET/MR compatible and do not produce visible artifacts or disturb image quality. The RT components showed a repositioning accuracy of better than 2 mm. Photon attenuation of \(-11.8\%\) in the top part of the phantom was observable, which was reduced to \(-1.7\%\) with AC using the \( \mu \)-map generator. Active lesions of 3 subjects were evaluated in terms of SUV\(_{\text{mean}}\) and an underestimation of \(-10.0\%\) and \(-2.4\%\) was calculated.
without and with AC of the RF body bridges, respectively. The new dedicated RT equipment for hybrid PET/MR imaging enables acquisitions in all body regions. It is compatible with PET/MR imaging and all hardware components can be corrected in hardware AC by using the suggested $\mu$-map generator. These developments provide the technical and methodological basis for integration of PET/MR hybrid imaging into RT planning.

Key words: PET/MR hybrid imaging, RT treatment planning, PET attenuation correction, PET quantification

(Some figures may appear in colour only in the online journal)

1. Introduction

In the last decade, radiotherapy (RT) treatment planning has conventionally relied on computed tomography (CT), especially in respect of dose calculations due to its accurate anatomical information and its direct relation to electron density. However, in modern RT treatment planning additional imaging modalities are essential for accurate tumor delineation. Magnetic resonance (MR) imaging with its excellent soft tissue contrast is often used as a complement reducing the risk for marginal tumor misses (Debois et al 1999). To overcome additional radiation by the CT and errors associated with image registration between CT and MR, several research groups have described the efficiency of MR-based RT treatment planning (Lee et al 2003, Karlsson et al 2009, Dowling et al 2012, Uh et al 2014). In recent years, positron emission tomography (PET), mostly in combination with CT has become an important additional imaging modality to improve target definition in RT treatment planning, because of its highly specific metabolic information (Ciernik et al 2003, Ford et al 2009, Nestle et al 2009).

Whole-body hybrid PET/MR imaging is a growing research field in medical imaging within the last years (Drzezga et al 2012, Quick et al 2013) potentially offering new possibilities for RT treatment planning (Thorwarth et al 2013), especially since PET/MR systems have become commercially available and are used in clinical routine now. An RT workflow scheme showing the potential benefits of using simultaneous PET/MR instead of separate MR and PET/CT is shown in figure 1. A first intensity-modulated radiotherapy (IMRT) treatment planning with $^{68}$Ga-DOTATOC-PET/MR has been performed and evaluated (Thorwarth et al 2011), but without using dedicated RT equipment that is necessary for reproducible patient positioning at all imaging modalities (Thorwarth et al 2012). Such equipment should include a flat table overlay, where the individually formed thermoplastic face mask and other positioning aids for reproducible patient positioning can be mounted at certain defined positions. Furthermore, MR imaging in RT treatment planning requires coil holders, which fix the otherwise flexible radiofrequency (RF) receiving coils at a predefined position. This is necessary, because the RF coils should not touch and deform the patient’s body as they do as surface RF coils in routine MR imaging. Dedicated RT equipment developed for PET/CT or MR imaging cannot be used on a PET/MR hybrid system, since their materials are not compatible with MR or PET imaging. First RT equipment for hybrid PET/MR imaging of the head/neck region has been introduced and tested in a previous study by Paulus et al (2014), however, the described RT setup is limited to head imaging only and at that time does not allow for fixation of the head, as it is required for RT treatment planning.

For excellent image quality in PET and accurate activity quantification, attenuation correction (AC) of the patient as well as of hardware components is necessary. Several groups have
evaluated the photon attenuation of different PET/MR hardware components and calculated significant PET signal loss that requires AC (Delso et al. 2010, Paulus et al. 2012). For patient tissues, AC in routine PET/MR hybrid imaging is based on MR-imaging based segmentation methods using an MR sequence (e.g., Dixon). This method allows for segmentation of up to four tissue classes including air, fat, lung, and soft tissue (Martinez-Mölle et al. 2009, Schulz et al. 2011). Attenuation correction of hardware components, which are placed in the PET field-of-view (FOV) is performed by using pre-acquired CT-based attenuation maps (μ-maps) that serve as 3D template models of the individual hardware components to be corrected. This method is only feasible, if the device is rigid and its position is fixed and known as it is the case e.g. for the PET/MR system’s patient table and RF receiving coils such as the RF spine array and the head/neck RF coil. Flexible RF coils are currently not included into PET/MR AC, since form and position of the coils vary between scans and the coil is not easily detectable with routine MR sequences. Different proposals have been made to include flexible RF coils into AC, such as using markers (Paulus et al. 2012, Kartmann et al. 2013, Eldib et al. 2014) or an ultrashort echo time (UTE) MR sequence (Paulus et al. 2012). However, these concepts have not been included into the routine reconstruction process of the PET/MR system yet.

In this work, new RT equipment for a hybrid PET/MR system enabling whole-body imaging has been developed and evaluated to demonstrate the technical and methodological feasibility of the integration of PET/MR into RT treatment planning. The RT equipment consist of a prototype flat RT table overlay, RF coil holders for head imaging, and RF body bridges for PET/MR body imaging that are tested towards PET/MR system integration. AC of all individual RT components was performed by generating 3D CT-based template models. A

Figure 1. RT planning workflow showing a series of steps from the diagnosis to the treatment delivery. In this scheme, dose calculations are based on the planning CT, but additional imaging information is used to improve the tumor delineation. Each modality requires its own RT components for appropriate patient positioning and all images have to be registered to the planning CT. The integration of hybrid PET/MR imaging (left, green) would reduce the number of RT components, acquisitions, and image registration steps compared to the combination of PET/CT and MR (right, red).
custom-built program for $\mu$-map generation is introduced that assembles all 3D AC templates depending on their presence and position in the PET field-of-view according to the RT setup during a patient examination. All RT components were evaluated in phantom experiments with regards to MR and PET imaging compatibility, attenuation correction, PET quantification, and position accuracy. The entire RT setup was then evaluated in a first PET/MR patient study on five patients with hybrid imaging at different body regions.

2. Materials and methods

2.1 PET/MR system

Imaging was performed on a whole-body hybrid PET/MR system (Biograph mMR, Siemens Healthcare GmbH, Erlangen, Germany), a 3 T MR system (gradient strength, 45 mT m$^{-1}$; and slew rate, 200 T m$^{-1}$ s$^{-1}$) with a fully integrated PET detector in its isocenter allowing for simultaneous PET and MR acquisitions (Quick et al 2013). The system is equipped with a set of RF receiver coils, including a 24-channel RF spine array coil, which is placed on top of the patient table, a 16-channel RF head/neck coil, and several flexible 6-channel RF body matrix coils.

2.2. RT components for PET/MR imaging

In the context of this work, prototype RT components (Qfix, Avondale, PA) have been designed and built that allow for dedicated RT patient positioning on a hybrid PET/MR system covering the entire body of the patient (see figure 2). The setup comprises a flat RT table overlay (length = 198 cm, width = 51 cm), two RF coil holders for head imaging, and two height-adjustable RF coil holders (body bridges) for body imaging applications. The RF coil holders are designed for the use of the standard flexible 6-channel RF body matrix coils. In combination with RT positioning aids, the setup allows for dedicated RT patient imaging on the PET/MR system from the head down to the feet depending on the region of interest.

The flat RT table overlay consists of a sandwich composite plastic with a foam core and is equipped with a Varian Exact® style indexing system, which has been evaluated in more detail by Paulus et al (2014). For this study pins for positioning a head/neck cushion, and small bore holes for fixing a thermoplastic head mask as seen in figure 2(a), were added to the RT table overlay. This allows for identical positioning of the patient’s head on the PET/MR system as on other imaging devices and during subsequent radiation treatment. The RT overlay is placed on top of the PET/MR system’s RF spine array coil and fixed at the head fixation system of the PET/MR scanner to provide exact and reproducible RT table positioning.

For head imaging, two RF head coil holders are used (figure 2(b)) each fixing a flexible 6-channel RF body receive coil that have been tested and systematically evaluated by Paulus et al (2014).

To extend the concept to whole-body PET/MR imaging for RT treatment planning, two additional RF body bridges have been designed in this work that each a receiving RF body matrix coil at any predefined $z$-position by using the indexing system of the RT table overlay as shown in figures 2(c) and (d). The RF body bridges are adjustable in two different heights allowing for individual positioning of the RF receiving coils depending on the patient size and the position of the region of interest in respect to the patient. Thus, the RF receiving coils are placed above the patient without touching and deforming the patient’s body. The RF body bridges can be arranged next to each other so the RF body matrix coils have a certain overlap
as shown in figures 2(c) and (d) allowing for coverage of two PET/MR bed positions. The overlap of the RF body matrix coils is also used in routine PET/MR imaging, when the RF coils are placed directly on the patient’s body to increase RF signal homogeneity across the MR FOV.

2.3. MR compatibility testing

All equipment that is used in conjunction with MR imaging needs to be tested for safety and compatibility with MR imaging. Different standards exist that define specific testing procedures for force and torque, RF heating, and image artifacts (Schaefers and Melzer 2006). The flat RT table overlay consists of a sandwich composite plastic with a foam core. The RF coil holders for head and for body imaging fully consist of plastic and do not contain any ferromagnetic or metallic components, nor do they contain electrical conducting materials such as carbon fibers. This reduces the necessary testing to evaluation of potential MR imaging artifacts (Schaefers and Melzer 2006).

In the context of this study all individual RT components were evaluated in phantom experiments with regards to their potential impact on magnetic field homogeneity and on artifact generation. The impact of RT components on the MR shim has been tested with a field sequence (0.25 ppm/line). Therefore, each RT device was placed next to a 24 cm diameter spherical oil phantom and each RT device was measured individually. The images were then...
compared to the reference scan without any RT device. Additionally, it was tested whether the components are visible in MR imaging when using fast gradient echo sequences with short echo time (TE = 1.79 ms).

2.4. RT component attenuation correction

Accurate AC of all hardware components that are potentially placed in the PET FOV is necessary to provide quantitative and qualitative PET images. Therefore, a CT-based $\mu$-map of each RT component was generated and its effect on PET activity quantification was evaluated. CT acquisitions of each component were performed on a stand-alone CT (Somatom Definition FLASH, Siemens Healthcare GmbH, Forchheim, Germany), all with 140 keV tube voltage. The patient table of the CT system and other aids used for proper component positioning were segmented out of the resulting CT data and the noise around the object was reduced by post-processing. The CT images were scaled to linear attenuation coefficients at 511 keV using the bilinear approach that is established for AC in PET/CT hybrid imaging (Carney et al 2006). Overall, 4 single RT hardware component $\mu$-maps were generated: RT overlay, RT head setup, and two RT body bridges (height 1 and 2).

2.5. $\mu$-map generator

The number of different hardware components used in conjunction with the suggested RT setup and their various individual positions in the FOV of the PET detector during simultaneous PET/MR data acquisition renders exact hardware attenuation correction of these components challenging. Overall, a large number of components and objects have to be considered during attenuation correction: the PET/MR system patient table, the patient tissues, the spine array RF coil, multiple flexible RF surface coils, and the suggested RT components such as the RT overlay, the RF coil holders for head imaging, and the body bridges that are all variable in their position. This leads to a complex situation and all individual components need to be considered with their accurate position and with their accurate attenuation properties in 3D space.

The PET/MR system’s patient table and rigid RF coils, such as the RF spine array coil, are automatically considered in the system’s reconstruction software during AC. If located in the PET FOV and depending on their position, 3D AC templates of these components are added to the hardware component $\mu$-map, which is used for AC together with the MR imaging-based human tissue $\mu$-map. Flexible and non-rigid RF coils are not included in AC, because their position in space and individual form can vary between scans. Flexible RF coils thus have been designed with very low attenuation (Paulus et al 2012) and position detection with markers has been suggested to consider flexible RF surface coils in the AC (Kartmann et al 2013).

When used in conjunction with the suggested RT components for PET/MR imaging, the flexible RF body matrix coils are either fixed in the rigid head setup at a certain position or in the rigid but height-adjustable RF body bridges at defined x-, y-, and z-positions.

In this work, an algorithm, referred to as $\mu$-map generator, was developed that reduces the complexity of the RT setup and ensures that all RT hardware components and the, otherwise, flexible RF surface coils in the PET-FOV are properly considered during AC. The program generates only one overall 3D $\mu$-map for the current RT setup that is then reported to the PET/MR system during PET data reconstruction.

A schematic overview of the $\mu$-map generator is shown in figure 3. Since the presence of the RT table overlay is not detectable by the PET/MR system and it cannot distinguish whether the RF body matrix coil is used for the head setup or the RF body bridges, the user has to provide information about presence and position of these devices.
The in-house written \(\mu\)-map generator program (IDL software, Exelis Visual Information Solutions, Inc., Boulder, CO) requests, whether the RT table overlay and the RT head setup are mounted during the PET/MR acquisitions and information are automatically added to the hardware component \(\mu\)-map if needed. Next, the user has to provide information about the RF body bridges. This includes the number (0, 1, or 2) of used RF body bridges, their individual height (two heights), orientation (head or feet direction), and position (along \(z\)-axis). While an individual \(\mu\)-map for each height of the RT body bridges is needed, neither the different positions along \(z\)-axis nor the orientations require an additional \(\mu\)-map. For the different orientations, the \(\mu\)-map is rotated 180° around the sagittal axis. The position along \(z\)-axis is changed by adapting the \(\mu\)-map offset depending on the index number of the flat RT table overlay, on which the RF body bridge is mounted (22 different index positions). Using all this information, an individual overall RT component \(\mu\)-map of any RT setup can be generated offline and used in the PET/MR system reconstruction process. Hereby, the \(\mu\)-map is added to the hardware component \(\mu\)-map of the PET/MR system (patient table and RF spine array coil) and used for attenuation and scatter correction together with the MR-based \(\mu\)-map of the patient.

2.6. Position accuracy

Automatic AC of all RT devices when using CT-based attenuation templates requires accurate positioning of each component for any acquisition setup. The RT table overlay is fixed in its position through the head fixation system of the PET/MR system and is thus not variable in its position. Positioning accuracy of the RT head setup was already tested (Paulus et al 2014). The RF body bridge setup was mounted and repositioned 8 times in this study. Hereby, the RF coils were also removed from the RF body bridges. Three active \(^{68}\text{Ge}\) rod sources were attached to the surface of the flexible RF coil as seen in figure 4(a) and PET acquisitions were
performed between each installation. The PET images (see figure 4(b)) were registered to each other and the standard deviation of the resulting registration was calculated for translation and rotation in order to evaluate the overall repositioning accuracy of these RT components.

2.7. PET/MR phantom acquisitions

Qualitative and quantitative PET evaluation with the RF body bridges was performed using a PET body emission phantom (PTW, Freiburg, Germany) according to NEMA standards. The phantom was filled with water and 289.6 MBq $^{18}$F, placed on top of the RT table overlay and two consecutive PET scans were performed for 13 min each. The first scan was performed without RF body bridges and used as standard of reference. In the second scan, one RF body bridge was attached to the RT table overlay covering the dimensions of the NEMA image quality (IQ) phantom. For all scans, attenuation correction of the phantom was performed using a CT-based $\mu$-map to take in consideration the phantom with its plastic housing and the exact linear attenuation coefficients for the water filling (Ziegler et al 2015). The system’s patient table and the spine RF coil were automatically included into the hardware component $\mu$-map for all phantom acquisitions. Individual $\mu$-maps for the RT components were assembled and combined using the described $\mu$-map generator. The RT table was added for all PET reconstructions. For the scan with the RF body bridge, two different $\mu$-maps were generated, one with the RF body bridge (AC) and one without (NoAC). Activity concentration differences to the reference scan were determined inside the phantom to calculate the attenuation of the RF body bridge. PET images were reconstructed iteratively using 3D ordinary Poisson ordered-subsets expectation maximization (OP-OSEM) with 3 subsets and 21 iterations Matrix size was set to $344 \times 344$ with $127$ slices, $(2.09 \times 2.09)$ mm$^2$ pixel spacing with 2.03 mm slice thickness.

2.8. PET/MR patient scans

The whole-body RT concept for PET/MR imaging has been clinically tested in an initial in vivo study on 5 patients (1 head scan, 4 body scans) to evaluate the later use of this RT equipment in PET/MR patient imaging. Hereby, workflow as well as quantitative and qualitative image quality has been evaluated. All patients underwent a clinically indicated PET/CT examination before being scanned on the PET/MR system, while no additional radiotracer was injected. The PET/MR acquisitions were started $151 \pm 29$ min after injection on average. All patients provided informed consent and the approval of the institutional review and ethical board was obtained. Patient information is listed in table 1.

The patients have been placed on the flat RT table overlay on top of the RF spine array coil for all scans and two PET/MR acquisitions of each patient have been performed. The first scan
was with the RF head coil holders (patient 1) or the RF body bridges (patient 2–5) depending on the bed position. For the second scan (reference) the respective RF coil holders were removed and the acquisition was started without moving the patient.

For each patient, a single PET bed position was acquired depending on the clinical diagnosis. Simultaneously, MR images were acquired for diagnosis and to provide the patient’s $\mu$-map. The human $\mu$-map was provided by the PET/MR system and is based on a coronal Dixon-VIBE (volumetric interpolated breath-hold examination) sequence with the following scan parameters: integrated parallel acquisition technique; factor, 2; $192 \times 126$ matrix with $2.6 \times 2.6 \text{ mm}^2$ pixel size, 128 slices a $3.12 \text{ mm}$, TR = $3.6 \text{ ms}$, TE1/TE2 = $1.23 \text{ ms}/2.46 \text{ ms}$, flip angle = $10^\circ$, 2D distortion correction, TA = $19$ s per bed. PET data of the patients were reconstructed iteratively using 3D OP-OSEM (3 subsets, 21 iterations) concerning to the standard reconstruction parameters used for PET/MR imaging: $344 \times 344$ with $(2.09 \times 2.09) \text{ mm}^2$ pixel spacing and 127 slices with $2.03 \text{ mm}$ slice thickness. PET images were post smoothed with a Gaussian filter of $4 \text{ mm}$ full width at half maximum.

Quantitative evaluation of the PET images was performed by comparing the standard uptake values (SUVs) of active lesions. Volumes of interest (VOIs) were drawn using a $50\%$ maximum contour of the PET SUV.

## 3. Results

### 3.1. Compatibility testing

During compatibility testing with an MRI visible phantom none of the RT components did show any influence on the magnetic field homogeneity and no other MR imaging artifacts or geometric distortions were observed. Furthermore, none of the RT hardware components showed any visible signal in MRI when imaging with a fast gradient echo sequence with short echo times.

### 3.2. Position accuracy

The standard deviation between all image registrations of the rod sources attached to the RF body bridges and the maximum deviations are listed in table 2. This includes left-right (LR), anterior-posterior (AP), and superior-inferior (SI) as well as all three rotation possibilities. The length of the calculated 3D vectors to the origin ranged from $0.60 \text{ mm}$ to $1.89 \text{ mm}$ (mean $1.29 \text{ mm} \pm 1.23 \text{ mm}$).

### Table 1. Patient characteristics.

<table>
<thead>
<tr>
<th>Patient</th>
<th>Age</th>
<th>Gender</th>
<th>Weight (kg)</th>
<th>Injected activity (MBq)</th>
<th>Bed position</th>
<th>Diagnosis</th>
</tr>
</thead>
<tbody>
<tr>
<td>01</td>
<td>26</td>
<td>Male</td>
<td>119</td>
<td>382</td>
<td>Head</td>
<td>Bronchial carcinoma</td>
</tr>
<tr>
<td>02</td>
<td>48</td>
<td>Male</td>
<td>70</td>
<td>299</td>
<td>Thorax</td>
<td>Bronchial carcinoma</td>
</tr>
<tr>
<td>03</td>
<td>61</td>
<td>Male</td>
<td>78</td>
<td>212</td>
<td>Thorax</td>
<td>Bronchial carcinoma</td>
</tr>
<tr>
<td>04</td>
<td>60</td>
<td>Male</td>
<td>97</td>
<td>224</td>
<td>Thorax</td>
<td>Bronchial carcinoma</td>
</tr>
<tr>
<td>05</td>
<td>54</td>
<td>Female</td>
<td>55</td>
<td>188</td>
<td>Feet</td>
<td>Synovial sarcoma</td>
</tr>
</tbody>
</table>

3.3. PET/MR phantom acquisitions

The percentage differences of the PET images with the RF body bridge (NoAC and AC) to the reference scan without the RF body bridge are shown in figure 5 for the central slice of the phantom. Additionally, 5 regions of interest (ROIs) have been drawn inside the phantom

### Table 2. Registration parameters calculated for the positioning accuracy of the RF body bridges.

<table>
<thead>
<tr>
<th>LR</th>
<th>AP</th>
<th>SI</th>
<th>Pitch (LR axis)</th>
<th>Yaw (AP axis)</th>
<th>Roll (SI axis)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.67 mm</td>
<td>0.45 mm</td>
<td>1.20 mm</td>
<td>0.34°</td>
<td>0.39°</td>
<td>0.18°</td>
</tr>
<tr>
<td>0.97 mm</td>
<td>0.84 mm</td>
<td>1.60 mm</td>
<td>0.54°</td>
<td>0.54°</td>
<td>0.40°</td>
</tr>
</tbody>
</table>

Figure 5. Phantom measurements with the body bridge and flexible RF coil placed on top of the NEMA IQ phantom. (a) and (b) Attenuation maps for the NEMA IQ phantom measurements separated in phantom/tissue $\mu$-map, hardware component $\mu$-map, and overall $\mu$-map (bottom). The $\mu$-maps are shown for the NoAC and the AC reconstruction of the NEMA scan. (c) Percentage difference of the central slice of the NEMA IQ phantom for NoAC and AC compared to the reference scan. 6 ROIs were drawn to evaluate local deviations as shown in the middle. Note increased PET signal attenuation in the upper parts of the NEMA phantom (left in c) due to the non-corrected body bridge and RF coil, but homogeneous low differences across the phantom (right in c) following attenuation correction of RF coil and coil holder. Note also that the flat RT table overlay was included into the overall $\mu$-map (a) and (b) for all acquisitions thus not showing additional attenuation bias from below the phantom.
Table 3. Percentage difference of the phantom measurements for each ROI compared to the reference scan.

<table>
<thead>
<tr>
<th></th>
<th>Top</th>
<th>Left top</th>
<th>Right top</th>
<th>Left bottom</th>
<th>Center</th>
<th>Right bottom</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>(1)</td>
<td>(2)</td>
<td>(3)</td>
<td>(5)</td>
<td>(4)</td>
<td>(6)</td>
</tr>
<tr>
<td>NoAC</td>
<td>−15.5% ± 2.8%</td>
<td>−10.0% ± 2.5%</td>
<td>−9.8% ± 2.7%</td>
<td>−4.5% ± 1.0%</td>
<td>−7.3% ± 2.3%</td>
<td>−4.0% ± 1.5%</td>
</tr>
<tr>
<td>AC</td>
<td>−3.7% ± 1.5%</td>
<td>−0.7% ± 1.5%</td>
<td>−0.8% ± 1.1%</td>
<td>−1.5% ± 1.0%</td>
<td>0.0% ± 1.7%</td>
<td>−0.6% ± 1.3%</td>
</tr>
</tbody>
</table>
and the percentage differences to the reference scan without the RF body bridges have been calculated (table 3) to assess local effects of the attenuation bias. For the ROIs close to the RF body bridges (ROIs 1–3) photon attenuation of $-11.8\% \pm 3.2\%$ was calculated in mean. With CT-based AC of the RT devices using the $\mu$-map generator, the bias was reduced to $-1.7\% \pm 1.7\%$ in mean. The attenuation effect was less than $-7.5\%$ in the center and the bottom ROIs (4–6), while with AC, the mean activity of the ROIs was almost equivalent to the reference scan (bias less than 2\%).

3.4. PET/MR patient scans

MR images of an axial 2D spin echo sequence are shown in figures 6(a) and (b) for the routine PET/MR setup and for the RT PET/MR setup, respectively. Both setups provided comparable MR image quality and signal homogeneity throughout the obtained FOV.

PET, MR, and superimposed PET/MR images of all 5 patients with the RT setup are shown in figure 7. Out of all 5 patients, only the abdominal patients (2–4) showed active lesions that where quantitatively evaluated towards SUV mean and SUV max as shown in table 4. Without the consideration of the RF body bridges in AC, a mean underestimation of $-11.1\% \pm 2.0\%$ and $-10.0\% \pm 2.4\%$ was calculated for the SUV max and SUV mean, respectively. The bias was reduced to $-3.9\% \pm 2.6\%$ (SUV max) and $-2.4\% \pm 3.3\%$ (SUV mean) with the CT-based AC of the RF body bridges.

4. Discussion

In this work, a whole-body concept has been developed and evaluated to integrate hybrid PET/MR imaging into RT treatment planning. The concept includes a flat RT table overlay, RF coil holders for head imaging, and RF body bridges for body imaging allowing for bed acquisitions from the head down to the feet. Compatibility with MR imaging has been tested with phantom measurements and the repositioning accuracy of the RT components has been evaluated, which is necessary for automatic PET AC. For each individual RT hardware component a CT-based 3D template model for attenuation correction was generated. A hardware component $\mu$-map was assembled using a custom-built $\mu$-map generator program, which combines the CT-based AC templates of all individual components depending on their
individual position and orientation within the PET FOV. The entire RT concept has been systematically evaluated on a PET/MR system with phantom scans and a first patient study of five subjects has been included.

The new RT equipment was evaluated towards its compatibility with MR imaging. Since neither ferromagnetic, nor metallic, nor electrical conducting materials were present in this prototype assembly, testing for potential MR imaging artifacts of all components was considered sufficient (Schaefers and Melzer 2006, ASTM F2119-07 Standard 2013) and did not
result in any measurable influence on MR imaging or in MR visible artifacts or structures. The prototype RT equipment thus can be considered MR compatible. Of course, a dedicated investigation on MR safety and compatibility testing along existing standards needs to be performed when medical product accreditation is pursued by the manufacturer.

The setup of the RT devices is straightforward and allows for body imaging from the head down to the feet. The flat RT table overlay consists of an indexing system for positioning of dedicated RT aids. The head part is equipped with pins, so a head/neck cushion can be placed on top of the RT overlay. Thermoplastic masks can be fixed using the bore holes in the RT table overlay. In combination with the RF coil holders for the head, appropriate RT imaging is feasible for the head/neck region. Single components for PET/MR imaging of the head in context with RT planning were already evaluated in a previous study by Paulus et al (2014), but the setup in that study did not allow for dedicated fixation of the head with masks and did not encompass body RF coil holders and an automatic AC method for an extension of the concept to whole-body PET/MR imaging.

For PET/MR body imaging, RF body bridges were designed to keep the flexible RF receiving coils at a fixed position above the body surface of the patient. The indexing system is used to attach the body bridges to the RT table overlay, which allows for translation in discrete steps in SI direction of the system. The RF body bridges are height adjustable in two steps to fit different patient sizes and to assure that the RF coils are close to the patient for optimized MR image quality while not touching or deforming the patients body.

Within this study, AC of the RT components was included into the process of PET reconstruction by using a \( \mu \)-map generator. Since none of the RT components has an RF plug that can be recognized by the PET/MR system, the \( \mu \)-map for the RT components has to be generated offline. Thus the user has to provide the information about the presence of each RT component to the software in the current implementation. Since the RT table overlay and the RT head setup have a fixed position, no further information is needed. The RF body bridges, however, can be placed at different \( z \)-positions, \( y \)-heights, and orientations. Thus, this information has to be added to the \( \mu \)-map generator. Due to the rigid form of the RF body bridges and attached RF coils, AC can be performed if the correct information is provided. An incorrect input might lead to a misregistration of the \( \mu \)-map to the actual position and thus lead

<table>
<thead>
<tr>
<th>Table 4.</th>
<th>( \text{SUV}<em>{\text{max}} ) and ( \text{SUV}</em>{\text{mean}} ) of the active lesions of patient 2–4 for the reference scan, NoAC, and AC.</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>( \text{SUV}_{\text{max}} )</td>
</tr>
<tr>
<td>Patient 2</td>
<td>Reference</td>
</tr>
<tr>
<td></td>
<td>NoAC</td>
</tr>
<tr>
<td></td>
<td>AC</td>
</tr>
<tr>
<td>Patient 3</td>
<td>Reference</td>
</tr>
<tr>
<td></td>
<td>NoAC</td>
</tr>
<tr>
<td></td>
<td>AC</td>
</tr>
<tr>
<td>Patient 4</td>
<td>Reference</td>
</tr>
<tr>
<td></td>
<td>NoAC</td>
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<td>AC</td>
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</tbody>
</table>

Note: Additionally, the percentage difference of NoAC and AC to the reference scan is listed.
to PET image artifacts or wrong PET quantification. To prevent the wrong orientation of the RF body matrix coil, the RF coil header of the PET/MR system might be used as additional information.

For exact AC it is necessary that the RT devices are repositioned accurately to match the virtual position of the 3D AC templates. This was assured by repeated PET acquisitions with radioactive rod sources. The registration between all images showed a maximum translation of less than 2 mm in all three directions, which is considered acceptable in the context of hardware AC (Paulus et al 2013). Repeated repositioning of RF body bridges and RF coils in this study has thus shown that this requirement is fulfilled.

The accurate repositioning allows for CT-based AC of all RT components. The calculated attenuation of the RF body bridges with the RF body matrix coil was $-11.8\%$ in the top part and $-5.3\%$ in the bottom part of the phantom, which is comparable to other studies (Kartmann et al 2013, Paulus et al 2013, Eldib et al 2014) where only a flexible RF coil was evaluated. The small difference between this study and the results from previous studies shows that the RF body bridges by itself only affect the PET signal a little.

When CT-based AC of the RT devices was performed, a remaining $-1.7\%$ attenuation in the top part of the phantom was observable, which might be due to the cable of the RF coil being disregarded in AC, since it remains as a flexible part of the setup. The results of the patient study confirmed the results of the phantom scan with $-10.0\%$ attenuation due to the RF body bridge with attached RF coil. The bias was reduced to $-2.4\%$ with AC using the $\mu$-map generator. The higher bias for patient 2 ($-12.3\%$ for NoAC and $-5.7\%$ for AC) is probably a result of using two attenuation corrected body bridges in the PET FOV and thus, the remaining (uncorrected) attenuation of two cables.

Acquisition of the feet was not performed with dedicated positioning aids, but with foam supports of the PET/MR system. RT positioning aids for knee and feet are available for MR and CT imaging by different vendors, but were not present for the PET/MR acquisitions. Such setup, however, is generally feasible, since the RT table overlay is equipped with a standard indexing system. Overall, the study shows that the presented RT setup enables PET/MR body acquisitions from the head down to the feet.

The prototype RT devices for hybrid PET/MR imaging of this study enable a reproducible positioning of the patient, which is needed throughout the whole imaging and treatment cascade. Thus PET/MR can be used as additional information to improve tumor delineation. Furthermore, it might improve the imaging workflow within the treatment, since instead of a PET/CT and a separate MR acquisition, only a PET/MR can be used additional to the planning CT as schematically shown in figure 1. Mostly, dose calculations are not performed based on the CT data of the PET/CT system, but of the planning CT (Thorwarth et al 2013). Since PET/MR images are acquired simultaneously, only one image registration to the planning CT has to be performed instead of two (PET/CT and MR), reducing potential registration errors. Moreover, it requires only one set of dedicated RT components, instead of one set for PET/CT and one for MR.

One limitation of this study is the low number of patients examined for the clinical evaluation. The main focus of this work, however, is on technical development, component testing, and system integration testing. The patient study in this context shows first examples of the feasibility of integrating PET/MR imaging into RT treatment planning in different body regions and thus provides the basis for PET/MR-based clinical studies. With the new RT equipment and its PET/MR attenuation correction method, larger patient studies can now be performed evaluating the benefits of integrating PET/MR into RT treatment planning.
5. Conclusion

Dedicated RT components have been developed and tested to enable whole-body hybrid PET/MR imaging in RT treatment planning. AC of all components was performed using a μ-map generator program that considers the presence and position of each RT component individually. Compatibility with PET and MR imaging was tested and the first PET/MR patient acquisitions were performed to demonstrate the feasibility of body imaging from the head down to the feet. These developments together with dedicated positions aids provide the basis for integration of PET/MR hybrid imaging into RT treatment planning.

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