A head coil system with an integrated orbiting transmission point source mechanism for attenuation correction in PET/MRI

Abstract
The combination of positron emission tomography (PET) and magnetic resonance imaging (MRI) provides a benefit for diagnostic imaging. Still, attenuation correction (AC) is a challenge in PET/MRI compared to stand-alone PET and PET-computed tomography (PET/CT). In the absence of photonic transmission sources, AC in PET/MRI is usually based on retrospective segmentation of MR images or complex additional MR-sequences. However, most methods available today are still challenged by either the incorporation of cortical bone or substantial anatomical variations of subjects. This leads to a bias in quantification of tracer concentration in PET. Therefore, we have developed a fully integrated transmission source system for PET/MRI of the head to enable direct measurement of attenuation coefficients using external positron emitters, which is the reference standard in AC. Based on a setup called the ‘liquid drive’ presented by Jones et al (1995) two decades ago, we built a head coil system consisting of an MR-compatible hydraulic system driving a point source on a helical path around a 24-channel MR-receiver coil to perform a transmission scan. Sinogram windowing of the moving source allows for post-injection measurements. The prototype was tested in the Siemens Biograph mMR using a homogeneous water phantom and a phantom with air cavities and a Teflon (PTFE) cylinder. The second phantom was measured both with and without emission activity. For both measurements air, water and Teflon were clearly distinguishable with air cavities and a Teflon cylinder. The linear attenuation coefficient was measured as \((0.096 \pm 0.005) \text{ cm}^{-1}\) in accordance with the true physical value. This combined MR head coil and transmission source system is, to our knowledge, the first working example to use an orbiting point source in PET/MRI and may be helpful in providing fully-quantitative PET data in neuro-PET/MRI.

1. Introduction

A unique feature of positron emission tomography (PET) is the sensitivity for tracer detection in the picomolar range (Pichler et al 2006). However, PET lacks the detailed anatomic information of magnetic resonance imaging (MRI) or computed tomography (CT). Efforts to physically combine PET and MRI can be traced back to the late 1980’s, but progress was slow at the beginning due to technological hurdles (Wehrl et al 2015). After significant progress in image co-registration (Hill et al 2001, Hutton et al 2002) of tomographic data from various modalities in the 1990’s, the development of a combined PET/CT system (Beyer et al 2000) has changed diagnostic nuclear medicine substantially. Hybrid imaging has become a standard in clinical oncological imaging. Nowadays, also PET/MRI technology has matured and started to enter the clinical arena (Partovi et al 2014). Initial patient studies and first clinical experience showed that PET/MRI is feasible (Drzegha et al 2012, Czermin et al 2014) and provides advantages compared to PET/CT in a number of clinical applications, for instance in head and neck cancer (Partovi et al 2014, Bashir et al 2015).

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Compared to stand-alone PET and PET/CT attenuation correction (AC) is still a challenge in PET/MRI (Keereman et al., 2013), especially for regions other than the brain. A detailed review of different methods for AC in PET/MRI is given amongst others by Izquierdo-Garcia and Catana (2016) and Ladefoged et al. (2017). Depending on the imaging modality used for obtaining the attenuation distribution the various approaches can be divided into MRI-based AC (MR-AC) and PET-based AC.

In the field of MR-AC we witnessed substantial progress over the last few years (Chen and An, 2017). Still, a unique global mapping function to convert MRI intensities to attenuation coefficients does not exist (Mehranian and Zaidi, 2015). Methods for MR-AC are typically segmentation-based or atlas-based. Methods relying on segmentation suffer from misclassification of tissue as well as assignment of fixed AC values. Especially cortical bone, which is an important tissue type in AC maps of the head, exhibits low signals on images acquired using conventional MR sequences. Thus, bone as well as air pockets, such as the sinuses cannot be well differentiated from each other for the generation of MR-AC maps. Therefore, the use of ultrashort-echo-time and zero-echo-time MR-sequences has been proposed (Keereman et al., 2010, Berker et al., 2012, Delso et al., 2015). However, some of these specialized sequences are sensitive to magnetic field inhomogeneities (Mehranian and Zaidi, 2015, Zaidi and Del Guerra, 2011). In addition, any variation of attenuation values is not considered in segmentation-based methods (Sekine et al., 2016).

Atlas-based MR-AC methods use a priori knowledge of the anatomy and corresponding attenuation properties. These atlas-based methods make use of co-registered MRI-CT datasets to derive a pseudo-CT image from the MRI data of a patient (Hofmann et al., 2008, Burgos et al., 2014, Mérida et al., 2017). In case of a perfect co-registration between atlas and patient data these methods are able to provide reliable MR-AC. However, perfect co-registration is rarely possible due to substantial anatomic variations among patients and the limitations of registration algorithms (Mehranian and Zaidi, 2015). Other challenging issues include invisible items to MRI, such as certain implants, contributing to photon attenuation as well as pathological and anatomical variations leading to significant deviations from the atlas (Delso et al., 2010, Zaidi and Del Guerra, 2011, Quick, 2014).

The second group of AC methods in PET/MRI is PET-based. In this case, the AC map is derived from the emission data alone (Salomon et al., 2011, Mehranian and Zaidi, 2015). Iterative reconstruction techniques such as the maximum likelihood reconstruction of activity and attenuation (MLAA) algorithm allow to reconstruct the activity distribution together with the attenuation properties. The performance of these methods can be improved using anatomical MRI information or by incorporating time of flight (ToF) information. However, these techniques do not provide unique solutions to the image reconstruction problem, thus, requiring at least a scaling of the results (Defrise et al., 2012). Moreover, these methods usually have problems in areas with low activity concentration such as patient boundaries or when using a tracer whose distribution is dominated by focal uptake. Both issues can be solved by the use of additional transmission information (Berker and Li, 2016).

There are already several groups working on incorporating a transmission source in PET/MRI. To date, the solutions published are either for ToF only or they do not support post-injection measurements. Simultaneous reconstruction of emission activity and attenuation coefficient distribution with an external transmission source was investigated in ToF PET/CT (Panin et al., 2013). Another approach uses the background radiation of the lutetium oxyorthosilicate (LSO) scintillators as a transmission source also in ToF PET/CT (Rothfuss et al., 2014). For ToF PET/MRI the use of an MRI-compatible positron-emitting source for simultaneous emission and transmission imaging was examined (Mollet et al., 2012, 2014). The authors used a toroidal transmission source filled with an aqueous solution containing $^{18}$F. The ToF information is utilized to separate emission and transmission data. For non-ToF PET/MRI fixed transmission source geometries were assessed (Bowen et al., 2016) using Monte Carlo simulations and an experimental setup of a single toroidal source. In addition, the use of transmission line sources to improve the estimation of truncated regions in the MR AC map was investigated (Watson, 2014).

The aim of our work is the development and validation of a setup using an orbiting point-like source that is fully integrated in an MR-receiver coil for head and neck measurements in non-ToF PET/MRI. The transmission source system hydraulically moves the point-like source on a helical path around the head coil. This idea is based on a setup called the ‘liquid drive’ (Jones et al., 1995). The AC map is calculated using the ratio of a blank and a transmission scan. The blank scan is usually acquired once a day. Using sinogram windowing (Carson et al., 1988) the transmission scan can be acquired simultaneously with the emission measurement. This should enable to obtain accurate AC maps in a manner that is compatible with the clinical workflow.

Transmission scans for the use of AC already have a long history in the case of stand-alone PET. Windowed rotating rod sources were used as standard geometry for transmission measurements (Bailey, 1998). With our work we want to revive this classic approach and introduce a method for transmission scans in PET/MRI using a point source. This enables the direct measurement of the attenuation factors for 511 keV photons without the need of making any assumptions on the object of investigation. Our approach is validated using phantom measurements and a comparison of the reconstructed linear attenuation coefficients with known values for the phantom materials.
2. Methods

We developed a head coil system for PET/MRI with a fully integrated orbiting point source to perform transmission scans. In section 2.1 the hardware of the developed prototype is described in detail. Section 2.2 elucidates the measurements to characterize this hardware in terms of possible scan-times and transmission source activity. Section 2.3 details the phantom study we used to validate our concept.

2.1. Hardware description

The head coil system consists of two parts: a hydraulic system for driving a positron-emitting point source on a helical path around the head of the subject being scanned and a custom-built 24-channel head and neck MR receiver coil. Both parts are fully integrated within one device. The prototype is designed for the Siemens Biograph mMR, however, it may be adapted for other platforms. The setup is shown in figure 1.

2.1.1. Hydraulic system

The hydraulic system is shown in figure 2. It consists of a tube connected to a centrifugal pump and valves to change flow direction (figure 2(c)) as well as a compensating reservoir (figure 2(b)). For initial tests purified water is used as hydraulic liquid. The tube is made of transparent polyurethane and has an inner diameter of 8 mm and an outer diameter of 12 mm. It is wound around the outer part of the housing of the head coil (figure 2(a)) in a helical fashion with a pitch of 12.25 mm. The diameter of a single winding is 320 mm and there are 20 windings in total, covering the entire 258 mm PET field of view (FoV) of the Siemens Biograph mMR system (Delso et al 2011). The flow direction can be reversed by combination of four 3-way valves. This adds flexibility to acquiring transmission scans. Changing the flow direction takes approximately 5 s. A compensating reservoir is inserted in the circuit to load and unload the transmission source. The reservoir, pump and valves are installed in the hydraulic system after a 7 m long section of the tube. This should allow for enough flexibility to cover the distance between the bore of the scanner and the waveguide for non-conductive connections of the PET/MRI room. The compensating reservoir is placed next to the waveguide inside the PET/MRI room. The pump and the valves are separated from the rest of the hydraulic system using self-sealing-couplings and are placed outside the PET/MRI room. Thus, the hydraulic system can be operated from outside the PET/MRI room. In addition, out of FoV activity can be avoided by moving the transmission source away from the scanner after a transmission scan. The self-sealing-couplings allow for quick installation and act as limit stop to prevent the pellet from leaving the PET/MRI room. For driving the source we installed a centrifugal pump (BP50 Pediatric Bio-Pump, Medtronic, Minneapolis, Minnesota, USA) with 1/4" inlet/outlet and 50 ml priming volume. The supply voltage can be continuously varied in the range of 0 V to 20 V DC to adjust the flow rate from 0 l min$^{-1}$ to 1.5 l min$^{-1}$. The performance of the pump in combination with the hydraulic system is characterized in more detail in sections 2.2.1 and 3.1.

2.1.2. Transmission source

The positron-emitter $^{18}$F is used as a source of 511 keV photons. An aqueous solution containing the positron-emitter is encapsulated within a refillable pellet (figure 1(b)). It is made of polyethylene with an overall length of 29 mm and an outer diameter of 7.4 mm. The cavity has a length of 14 mm and a diameter of 6 mm leading to a holding capacity of 0.4 ml. The cavity may also be filled with any other tracer required.

2.1.3. MR receiver coil

The custom-built head/neck coil consists of 24-channels distributed in four rows with 3/7/7/7 elements. Within each row, transformer decoupling was implemented; neighboring rows are decoupled by coil overlap. In addition, all elements are decoupled by preamplifier decoupling. Radiodense components like preamplifiers are placed outside the PET FoV. Further information about the coil as well as reconstructed MR images acquired with the coil can be found in Navarro de Lara et al (2017).

2.1.4. Housing

The housing is 3D-printed from laser-sintered polylaurinlactam, which is a biocompatible material. It consists of an elliptical inner and a cylindrical outer shell. The MR coil elements are attached to the inner shell. The elliptical shape allows for a minimum distance of the coil elements to the patient’s head. The outer shell acts both as cover for the MR coil and as support for the tube of the hydraulic system. To keep the windings of the tube at a fixed position four guide bars are attached to the outer part of the housing, at intervals of 90° (bottom, left, top, right). The wall thickness of both the inner and outer shell is 3.5 mm. The pedestal of the housing is fit to the shape of the patient table to ensure reproducible positioning of the head coil system.
2.1.5. Positioning of transmission source
An optical trigger system (figure 3) with 16 channels is utilized for the localization of the transmission source for measurements to characterize the hydraulic system described in section 2.2. Each channel consists of a binary optical trigger and a pair of polymer optical fibers together with an LED and a photo-diode. Each pair of fibers is connected to the liquid drive using drilled holes in the guide bars. There are four guide bars and 20 windings. Thus, there is a total of $81 \times (4 \times 20 + 1)$ possible measurement positions that the 16 optical triggers can be connected to. The light of the LED is guided to the tube of the liquid drive using one fiber. Each pair of holes is arranged along the flow direction having a distance of 4 mm. If there is no transmission source present at the respective position the light is reflected by the housing. The reflected light is guided back to the binary optical trigger using the other fiber. The detected signal is compared to a threshold value. If the transmission source passes a pair of fibers, the light is blocked and the detected signal falls below the threshold value. In this case the channel number together with the corresponding time-stamp is transmitted to a PC via USB. Thus, the position of the transmission source can be determined at 16 discrete locations for a given arrangement of the optical fibers (details about the arrangement are given in section 2.2).

2.1.6. Sinogram windowing
Sinogram windowing allows for separation of transmission and emission data and reducing both scatter and random coincidences in the transmission data. A sinogram window is basically a segmentation of the...
transmission source in a sinogram of a small time fraction of the whole acquisition. To obtain a sinogram window, each transmission scan is divided into 500 time frames using the listmode data of the PET system. Thus, each frame corresponds to an acquisition time of typically 160 ms and a distance along the tube of $\sim 4$ cm for a 80 s transmission scan. The sinogram window was chosen taking into account the size of the source (1.4 cm), the maximum positron range of the nuclide used (2 times 0.24 cm), as well as the transverse tangential resolution of the scanner as the position of the pellet is obtained from the PET data. Due to the high specific activity of the transmission source the source position can be segmented in sinogram space to obtain a sinogram window (figure 4). This is still true for the case of emission activity being present in the acquisition. Counts outside the sinogram window (red lines in figure 4(b)) are rejected for the transmission data.

To account for speed differences of the transmission source among scans, start and end time of each time window is adjusted to maximize the overlap of respective sinogram windows using the first blank scan as reference. This is implemented by measuring the difference in pixel between corresponding sinogram windows in sinogram space and converting it to a time difference considering sinogram size and transmission scan-time. The time difference is used to get the updated sinogram window from listmode data.

2.2. Characterization of hardware-performance

2.2.1. Hydraulic system—scan-time

The flow rate is affected by the power of the pump and the dimensions of the hydraulic circuit. The range of transmission scan-times achievable with our hydraulic system is determined by measuring the scan-time for different supply voltages. As scan-time we define the time the transmission source takes to move through all 20 windings of the tube. For the measurement only two optical triggers were used: one at the beginning of the windings (measurement position 1) and one at the end of the windings (measurement position 81).

2.2.2. Hydraulic system—stability of flow rate

To ensure the transmission source is moving at a constant speed, we investigated the motion for different overall densities of the pellet and different flow rates. A different density is achieved by varying the filling level of purified water in the pellet between 0.2 ml and 0.4 ml with a step-size of 0.05 ml. At a filling level of 0.3 ml the pellet is neutrally buoyant. The speed was varied in the full range feasible with our pump. The temperature of the setup was 24 °C. The consistency of the flow rate was evaluated by comparing the resulting speed of the pellet in all four segments of a winding. For the measurement optical triggers were placed at three consecutive windings in the middle of the head coil (measurement positions 33–45) as well as at the beginning and end of the 20 windings (measurement position 1 and 81) to compare the speed in a segment to the average speed in the whole system. A mean difference between segments below 1% is regarded as consistent flow rate.

2.2.3. Transmission source—optimal activity

To assess the optimal activity of the transmission source the noise equivalent count (NEC) rate of our system was examined. The NEC rate is given by (Strother et al 1990)

$$NEC = \frac{T^2}{T + 2R + S}$$

with true (T), random (R) and scattered (S) coincidences. Here, the factor two applies, because R is estimated from a delayed coincidence window. To obtain true and random coincidences count rate curves two measurements were performed on the Siemens Biograph mMR. The first measurement investigates the case of a post-injection

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**Figure 3.** Schematic of optical trigger system. The light of an LED is guided to the tube of the liquid drive. If there is no pellet the light is reflected and detected by the photodiode. In case a passing pellet blocks the reflection the detected signal is lower than the reference voltage of the microcontroller. A time-stamp of this event is then transmitted to a PC via USB.
transmission scan with high activity in the PET FoV. A cylindrical water phantom (height: 128 mm, diameter: 100 mm) filled with 1000 ml of water and an initial activity of 40 MBq (FDG) was placed inside the head coil to mimic a signal arising from a patient/object. The second measurement used the same water phantom without any activity to approximate the behavior for pre-injection transmission scanning. In both cases, a point source with a maximum initial activity of 500 MBq was placed on top of the head coil system in the center of the PET FoV. To acquire the count rate curves 1 min scans were acquired over a period of 90 mins. To expand the range of transmission source activity of this measurement three different point sources (500/360/180 MBq) were prepared and consecutively placed on top of the head coil over the period of 90 min. As scatter-fraction five different values (10/20/30/40/50%) were used to approximate the behavior of the NEC rate for different scattering bodies.

2.3. Phantom study (cold and hot)
The head coil system was validated in the Siemens Biograph mMR using two different phantoms. Both phantoms (figure 5) were manufactured in-house to fit the size of our head coil. Both have a cylindrical shape (inner diameter: 172 mm; height: 204 mm; wall thickness: 4 mm) and are made of polymethylmethacrylate (PMMA).

All activity values are determined using a cross-calibrated well counter and are decay corrected to mid acquisition time.

2.3.1. Phantom I
It is a homogeneous water phantom (figure 5, (a)) with a total filling volume of 4.75 l.

2.3.2. Phantom II
It is a phantom with fillable rods (figure 5, (b)) with a filling volume of the main cylinder of 4.2 l. The phantom contains a polytetrafluoroethylene (PTFE) cylinder with a diameter of 26 mm in the central axis. Around this cylinder there are six fillable rods made of PMMA arranged in three opposed pairs with an inner diameter of 26 mm, 9.5 mm and 4.5 mm (wall thickness: 2.2 mm, 1.85 mm and 1.2 mm respectively).

2.3.3. Cold measurement
Both phantoms were filled with purified water. The fillable rods of Phantom II were air-filled. The pellet was filled with 0.3 ml of an aqueous solution containing $^{18}$F. The activity was 76 MBq for the scan of Phantom I and 83 MBq for the scan of Phantom II. To achieve consistent flow rate of the pellet (compare section 3.1.2) the liquid drive was operated with a supply voltage to achieve a scan-time for a single transmission scan of around 85 s. For the acquisition of Phantom I seven transmission scans were performed resulting in a total scan-time of 10 min. For the acquisition of Phantom II six scans with a total time of 8.5 min were performed. Prior to the transmission scans an acquisition of 15 blank scans with a total time of 21 min was recorded with a transmission source activity of 96 MBq.

2.3.4. Hot measurement
To evaluate the performance of our method for the case of post-injection measurements, Phantom II was additionally measured with emission activity. The main cylinder was filled with $^{18}$F activity of 7 kBq ml$^{-1}$. Three rods (one of each pair) with a total volume of 160 ml were filled with an activity of 53 kBq ml$^{-1}$. The activity of the transmission source was 57 MBq. The total acquisition time was 18 min (3 transmission scans + 6 min emission only + 4 transmission scans).
2.3.5. Reconstruction of AC map

All acquisitions were corrected for decay of the transmission source and the number of scans performed. The first blank scan was used as reference to generate sinogram windows (see section 2.1.6). Start and end time of respective sinogram windows of the remaining blank scans and all corresponding transmission scans were adjusted to maximize for overlap of the sinogram windows. Sinograms were generated from listmode data using the prompt minus the delayed event packets. To reduce data and to improve the signal-to-noise ratio the single-slice rebinning (SSRB) algorithm (Daube-Witherspoon and Muehllehner 1987) was used to create rebinned direct sinograms. This introduces axial blurring with increasing distance from the axial center of the scanner. As we operate only in the head region and do not need to use the full transaxial FoV of the scanner the resolution loss is acceptable. For both blank and transmission scan all sinogram windows are added up to obtain the total blank and the total transmission sinogram. These sinograms have no gaps as the whole PET FoV is covered by the transmission source in the course of a scan. Prior to reconstruction both transmission and blank sinograms were smoothed with a $3 \times 3 \times 3$ Gaussian filter to reduce noise. For the case of post-injection the acquisition-time-corrected emission-only-sinogram was subtracted from the transmission sinogram to avoid emission contamination of the transmission data. To obtain the AC map the logarithm of the ratio of blank and transmission sinograms was calculated. This represents the integral of the attenuation coefficients along each LOR. To further reduce noise a $3 \times 3 \times 3$ Gaussian filter is again applied after calculating the log-ratio. This logarithm of the ratio was reconstructed using an ordered-subset expectation maximization (OSEM) algorithm (Hudson and Larkin 1994) for visualization and to obtain the linear attenuation coefficient (LAC) of each voxel. For the OSEM reconstruction 12 subsets and 24 iterations are used. The size of the image matrix is $344 \times 344 \times 127$ voxels. The advantage of the OSEM reconstruction over filtered backprojection were reduced streak artifacts as well as a better SNR in regions with low transmission count rate.

To evaluate the influence of total transmission scan-time the AC maps are reconstructed using one/two/all transmission scans of each acquisition.

2.3.6. Reconstruction of activity distribution

The AC map obtained with the transmission scan is used to reconstruct the emission activity of the hot measurement together with the transmission activity. As emission and transmission activity are spatially separated the transmission activity can be easily removed from the reconstructed activity distribution using a mask as a final post-processing step. The result is compared to the reconstructed activity distribution using AC map of the standard Dixon MR sequence of the PET/MRI scanner. The AC map of the head coil system itself is generated using Carney bilinear scaling (Carney et al 2006) of a CT of the whole setup. The activity distribution is reconstructed using the vendors e7-tools for mMR, which also incorporates scatter correction.

2.3.7. Evaluation of results

The numerical values of the reconstructed AC map are compared to the true theoretical and experimental attenuation coefficients (Hubbell 1969) of the respective materials. The value for PTFE is taken from the NIST database (National Institute of Standards and Technology; U.S. Department of Commerce). In these published tables the mass attenuation coefficient is given in cm$^2$ g$^{-1}$ for a photon energy of 500 keV and 600 keV. Linear interpolation is used to derive a reference value for 511 keV. Considering the density of the materials at room temperature the reference values for the different materials used in our phantoms are listed in table 1.
3. Results

3.1. Performance of hydraulic system

3.1.1. Scan-time

The range of feasible transmission scan-times is 30 s (supply voltage of pump: 17 V) to 20 min (2 V). Within this range the scan-time is continuously adjustable. Above a supply voltage of the pump of 17 V bubble formation in the hydraulic system is observed, below 1.8 V the transmission source does not move.

3.1.2. Consistency of flow rate

Consistent flow of the transmission source was observed for scan-times shorter than 90 s for all investigated filling levels of the pellet. For filling levels between 0.25 ml and 0.35 ml the relative difference between segments was up to 2% for scan-times of 5 min and up to 4% for scan-times of 15 min. For a filling level of 0.2 ml and 0.4 ml the relative difference between pellet speeds in segments was up to 3% for scan-times of 5 min and up to 10% for scan-times of 15 min.

3.2. Transmission source

3.2.1. Optimal source activity

The scatter fraction did not influence at which activity of the transmission source the peak NEC rate is reached. For the pre-injection case the peak NEC rate is reached with a transmission source activity of 158 MBq. Adding emission activity, the peak NEC rate is reached with a transmission source activity of 132 MBq (activity in cylindrical phantom: 27 MBq; total activity: 159 MBq).

3.3. Phantom study

3.3.1. Cold measurement

The reconstructed AC map of Phantom I and corresponding line profiles are shown in figure 6 for three different transmission scan-times; the position of the line profile is marked with a bright line in the reconstructed AC map. The top row shows the result using all seven transmission scans of the acquisition; the middle row using two scans; the bottom row using one scan.

Table 1. Reference values for the linear attenuation coefficient (LAC) at 511 keV.

| Material | Density (g cm\(^{-3}\)) | LAC (cm\(^{-1}\)) |
|----------|--------------------------|------------------|
| Water    | 0.9982                   | 0.0958           |
| Air      | 0.0012                   | 0.0001           |
| PMMA     | 1.18                     | 0.1101           |
| Teflon   | 2.2                      | 0.1838           |

Figure 6. Reconstructed AC map of Phantom I (cold) and corresponding line profiles for different transmission scan-times; the position of the line profile is marked with a bright line in the reconstructed AC map. The top row shows the result using all seven transmission scans of the acquisition; the middle row using two scans; the bottom row using one scan.
The reconstructed AC map of Phantom II and corresponding line profiles are shown in figure 7 for three different transmission scan-times. Reconstructed LAC values are reported in table 3 for six ROIs containing water, two ROIs for air and one ROI for Teflon. The position of all ROIs is shown in figure 7 (bottom row, right). For the large air rod as well as the Teflon cylinder the cylindrical ROI has a diameter of 1 cm to minimize the influence of partial volume error.

### 3.3.2. Hot measurement

The reconstructed AC map of Phantom II and corresponding line profiles are shown in figure 8 for three different transmission scan-times. Reconstructed LAC values are reported in table 3 for six ROIs containing water, one ROI for air and one ROI for Teflon. The position of all ROIs is shown in figure 7 (bottom row, right). For the large air rod as well as the Teflon cylinder the cylindrical ROI again has a diameter of 1 cm.

### 3.3.3. Reconstructed activity distribution

The reconstructed activity distribution using both the standard MR-AC map of the scanner and our proposed method are shown in figure 9. For the MR-AC map the activity was underestimated both for the background with \((5.0 \pm 0.5)\) kBq ml\(^{-1}\) and the big hot rod with \((39 \pm 3)\) kBq ml\(^{-1}\). Using the AC map obtained with the transmission scan yielded a more accurate activity distribution (background: \((6.8 \pm 0.7)\) kBq ml\(^{-1}\); big hot rod:

### Table 2. Reconstructed LAC values for different transmission scan-times of Phantom I (homogenous water phantom); reference value for water: \(0.096\) cm\(^{-1}\).

| tx scan-time (s) | LAC (cm\(^{-1}\)) |
|------------------|------------------|
| 79               | \(0.065 \pm 0.011\) |
| 158              | \(0.096 \pm 0.009\) |
| 554              | \(0.096 \pm 0.005\) |

![Figure 7. Reconstructed AC map of cold measurement of Phantom II and corresponding line profiles for different transmission scan-times; the position of the line profile is marked with a bright line in the reconstructed AC map; in the line profiles the reference values for water (LAC water) and Teflon (LAC Teflon) are plotted with a dashed line. The top row shows the result using all six transmission scans of the acquisition; the second row using two scans; the third row using one scan. The bottom row shows a CT of Phantom II to illustrate the orientation of the phantom during the measurement; in addition, on the right side the ROIs used for the evaluation of LAC values are shown.](image-url)
In addition, using the MR-AC method gives the incorrect geometry (inner diameter of phantom: 160 mm; diameter of PTFE cylinder: 37 mm) whereas the transmission scan gives the correct geometry (170 mm; 27 mm). For the transmission scan the LAC of air is overestimated (compare table 3). As a result the reconstructed activity for the big cold rod is overestimated (compare figure 9, top row).

4. Discussion

In this work we present a prototype head coil system to enable post-injection transmission measurements in non-ToF PET/MRI. In addition to post-injection scanning, the improved geometric windowing allowed by a point source compared to a line source reduces both random and scatter fraction and hence produces more accurate LACs even for pre-injection transmission scans. The injected activity distribution could be successfully reconstructed using the AC map obtained from the transmission scan. The MR-AC method failed mainly because of the misclassification of PTFE as air as this sequence is not designed for this kind of measurement. This also shows the strength of the transmission method: it does not need any assumptions about the object/patient being imaged.

The true LAC of water is successfully reproduced in phantom studies, both for transmission-only measurements and post-injection transmission measurements. To our knowledge we are the first group to perform post-injection transmission scans with a moving source in non-ToF PET/MRI. Comparison of the cold and the hot scan of Phantom II did not show substantial changes of the AC map due to the present activity. However, this was only possible after emission subtraction. Without subtraction the LAC is underestimated in regions with activity. Both hot and cold measurement yielded a considerably bigger LAC for air inside the rods (∼0.03 cm⁻¹; compare table 3) than the true value as well as a smaller value for the PTFE cylinder. Outside the phantoms the LAC of air is well below 0.01 cm⁻¹ (0.003 ± 0.001 cm⁻¹). The difference to the reference standard for the air rods and PTFE can be explained with partial volume error and with scattered coincidences within the sinogram window as the discrepancy to the reference standard became smaller using tighter sinogram windows as well as by reducing the ROI size. This issue has to be addressed in future work.

Table 3. Reconstructed LAC values of cold and hot measurements for different transmission scan-times of Phantom II; reference value for water: 0.096 cm⁻¹ and Teflon: 0.184 cm⁻¹; deviation of LAChot from LACcold is given as % difference.

| tx scan-time | ROI   | LACcold (cm⁻¹) | LAChot (cm⁻¹) | % difference |
|--------------|-------|----------------|---------------|--------------|
| 79 s (cold)  | Water₁ | 0.075 ± 0.008  | 0.064 ± 0.008 | -14.7        |
|              | Water₂ | 0.076 ± 0.010  | 0.067 ± 0.009 | -11.8        |
|              | Water₃ | 0.072 ± 0.008  | 0.059 ± 0.008 | -18.1        |
| 80 s (hot)   | Water₄ | 0.069 ± 0.009  | 0.059 ± 0.008 | -14.5        |
|              | Water₅ | 0.072 ± 0.009  | 0.057 ± 0.009 | -20.8        |
| 160 s (cold) | Water₁ | 0.082 ± 0.007  | 0.079 ± 0.008 | -3.7         |
|              | Water₂ | 0.081 ± 0.008  | 0.080 ± 0.008 | -1.2         |
|              | Water₃ | 0.081 ± 0.007  | 0.076 ± 0.007 | -6.2         |
| 158 s (hot)  | Water₄ | 0.079 ± 0.007  | 0.077 ± 0.008 | -2.5         |
|              | Water₅ | 0.081 ± 0.008  | 0.074 ± 0.009 | -8.6         |
| 475 s (cold) | Water₁ | 0.095 ± 0.005  | 0.096 ± 0.004 | +1.1         |
|              | Water₂ | 0.094 ± 0.005  | 0.096 ± 0.004 | +2.1         |
| 553 s (hot)  | Water₄ | 0.096 ± 0.005  | 0.095 ± 0.005 | -1.0         |
|              | Water₅ | 0.096 ± 0.006  | 0.097 ± 0.004 | +1.0         |
|              | Water₆ | 0.097 ± 0.006  | 0.096 ± 0.005 | -1.0         |
| 54 ± 4 kBq ml⁻¹) In addition, using the MR-AC method gives the incorrect geometry (inner diameter of phantom: 160 mm; diameter of PTFE cylinder: 37 mm) whereas the transmission scan gives the correct geometry (170 mm; 27 mm). For the transmission scan the LAC of air is overestimated (compare table 3). As a result the reconstructed activity for the big cold rod is overestimated (compare figure 9, top row).
The phantom study also revealed a significant negative bias for the LAC values as the transmission scan duration decreases. This bias is worse for the hot measurement compared to the cold measurements. With shorter transmission scan time the number of LORs with zero counts in the transmission scan increases. As a result, the ratio of the blank and the transmission scan is not defined for such a LOR. A meaningful value can still be assigned using the information from the neighbouring pixels, which potentially contain scattered events. The previously empty LOR did not contain any scattered events. Thus, scatter is amplified, which leads to the decrease of the LAC value. For hot measurements the effect is stronger due to the emission subtraction.

For our prototype transmission source system, a consistent flow was achieved using scan-times of less than 90 s. A longer transmission scan-time can still be achieved by reversing the flow direction and performing several scans within a single PET acquisition. In addition, if the specific gravity of the transmission source is matched to the density of the surrounding liquid, the influence of gravity is avoided and an almost uniform motion (within 2% deviation) of the transmission source can also be achieved for scan-times longer than 90 s.

The typical total time of a transmission scan is well below 10 min even for the case of a transmission source activity of less than 100 MBq. Due to sinogram windowing the transmission scan can be acquired post-injection to not interfere with clinical routine. However, changing the flow direction and monitoring the entrance/exit of the transmission source to/from the PET scanner should be automated for clinical routine by adding sensors checking for the passage of the transmission source and adding microprocessor controlled 3-way-valves.

As a final protocol we envisage a 30 min blank scan using a transmission source activity of 160 MBq (compare result of pre-injection NECR measurement in section 3.2) once a day and a 9 min transmission scan consisting of six 90 s scans using a transmission source activity of 130 MBq (compare result of post-injection NECR measurement in section 3.2). In the case of using a $^{68}$Ga/$^{68}$Ge point source where most likely just one source is available the blank scan can also be performed with 130 MBq and a longer acquisition time. Between each of the six transmission scans there should be a period of emission-only acquisition. This allows for emission subtraction in the transmission data and accommodates changing emission distribution over time.

For most PET/MRI acquisitions of the head the emission activity present in a patient will be well below the activity at peak NECR of the scanner (compare Delso et al (2011)). Thus, the additional activity of the trans-
mission source does not compromise the count rate performance of the PET system. For measurements with emission activity close to the activity at peak NECR the additional activity of the transmission source should be avoided by performing a pre-injection transmission scan.

In the original version of the ‘liquid drive’ developed by Jones et al. $^{137}$Cs was used as transmission point source. Since this radionuclide is not a positron emitter, but a single-gamma emitter it would require a modification of the PET/MRI system, more precisely of its coincidence processor, which is not possible in existing commercial devices. Thus, positron emitters are easier to implement routinely. To date, the transmission source is a small-sized pellet that is filled with $^{18}$F. It is cost-effective and widely available. The activity can be adapted for each experiment. However, the half-life is too short and the filling of the pellet too cumbersome for routine applications. The refillable pellet can be replaced by a coated $^{68}$Ga/$^{68}$Ge point source. This source could be manufactured smaller than the existing pellet. Thus, geometrical windowing could be further improved. The coating of the $^{68}$Ga/$^{68}$Ge source should seal the radionuclide from the environment and ensure neutral or near neutral buoyancy for consistent flow.

For the hypothetical clinical neuro application the additional patient dose in a head PET/MRI study caused by the transmission measurement was already quantified by several researchers in the past (Almeida et al. 1998, Wu et al. 2004). Reported values would translate to an ED of 50 μSv for our setup considering the smaller distance between source and patient, and using a 150 MBq transmission source for a 12 min scan. Thus, the issue of added radiation dose represents no limitation to the use of the presented transmission source system.

The liquid drive technology could be implemented into head-only PET systems such as the brain PET scanner (BPET) or preclinical PET-only systems for small animal measurements to provide a cost-efficient method for attenuation correction and avoid the need for an additional CT scan. For the use in whole-body scanners the liquid drive concept would require a much higher speed of the transmission source to achieve sufficiently short transmission scan times. Further, given the larger diameter of the windings required for whole-body applications would add additional material in the PET FoV. To account for these issues, the transmission source system could be altered in a way to use a cantilever with a single-gamma-emitting point source which can be moved across the PET FoV to acquire a
transmission scan. Thus, there would be less added material in the PET FoV. In addition, these gamma emitters have the advantage of achieving higher count rates and hence less noise in the attenuation map. As mentioned before, such a source would require a modification of the coincidence processor of the PET scanner as a line of response is obtained from the source position and single events at opposed detectors. On the other hand, using $^{137}$Cs with a photon energy of 662 keV would also allow for separation of transmission and emission data using a dual energy window.

In a next step post-injection in vivo measurements in piglets are planned before ultimately testing the prototype in humans which requires CE certification or similar. For future studies the purified water of the hydraulic system will be replaced by heavy water to avoid interference of the MR signal.

5. Conclusion

We successfully demonstrated the feasibility of post-injection transmission scans for head PET/MRI using an orbiting positron-emitting point-like source and presented a fully functional prototype for the Siemens Biograph mMR.

Acknowledgments

The authors thank Ewald Unger and the mechanical facility team of our centre for manufacturing of the pellet and the compensating reservoir. We thank Francesco Moscatò and the group of cardiovascular dynamics for their support in building the hydraulic system and the phantoms.

The financial support of the Austrian Science Fund (FWF) Project P27853-B30 is gratefully acknowledged.

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