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**A dual-tuned $^{13}\text{C}/^1\text{H}$ head coil for PET/MR hybrid neuroimaging:
Development, attenuation correction, and first evaluation**

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Abstract

Purpose

This study aims to develop, implement, and evaluate a dual-tuned $^{13}\text{C}/^1\text{H}$ head coil for integrated positron emission tomography/magnetic resonance (PET/MR) neuroimaging. The radiofrequency (RF) head coil is designed for optimized MR imaging performance and PET transparency and attenuation correction (AC) is applied for accurate PET quantification.

Material and Methods

A dual-tuned $^{13}\text{C}/^1\text{H}$ RF head coil featuring a 16-rung birdcage was designed to be used for integrated PET/MR hybrid imaging. While the open birdcage design can be considered inherently PET transparent, all further electronic RF components were placed as far as possible outside of the field-of-view (FOV) of the PET detectors. The RF coil features a rigid geometry and thin-walled casing. Attenuation correction of the RF head coil is performed by generating and applying a dedicated 3D CT-based template attenuation map (μmap). Attenuation correction was systematically evaluated in phantom experiments using a large-volume cylindrical emission phantom filled with 18-F-Fluorodesoxyglucose (FDG) radiotracer. The PET/MR imaging performance and PET attenuation correction were then evaluated in a patient study including six patients.

Results

The dual-tuned RF head coil causes a mean relative attenuation difference of 8.8% across the volume of the cylindrical phantom, while the local relative differences range between 1% and 25%. Applying attenuation correction, the relative difference between the two measurements with and

without RF coil is reduced to mean value of 0.5%, with local differences of $\pm 3.6\%$. The quantitative results of the phantom measurements were corroborated by patient PET/MR measurements. Patient scans using the RF head coil show a decrease of PET signal of $5.17\% \pm 0.81\%$ when compared to the setup without RF head coil in place, which served as a reference scan. When applying attenuation correction of the RF coil in the patient measurements, the mean difference to a measurement without RF coil was reduced to $-0.87\% \pm 0.65\%$.

Conclusion

A dual tuned $^{13}\text{C}/^1\text{H}$ RF head coil was designed and evaluated regarding its potential use in integrated PET/MR hybrid imaging. Attenuation correction was successfully applied. In conclusion, the RF head coil was successfully integrated into PET/MR hybrid imaging and can now be used for $^{13}\text{C}/^1\text{H}$ multinuclear hybrid neuroimaging in future studies.

Keywords: PET/MR hybrid neuroimaging, $^{13}\text{C}/^1\text{H}$ multinuclear, RF head coil, attenuation correction, ^{13}C hyperpolarization

Introduction

Hybrid imaging with simultaneous positron emission tomography and magnetic resonance (PET/MR) has demonstrated to be a powerful diagnostic modality for various applications in . While MRI provides excellent soft tissue contrast and various tissue parameters such as diffusion and perfusion, PET provides diagnostic information about tissue function and allows for accurate quantification of radiotracer distribution in the human brain and entire body .

Simultaneous image acquisition in PET/MR requires the use of radiofrequency (RF) coils for MRI signal detection that are optimized regarding their PET transparency. Conventional RF coils for MR-only use are often not well suited for the operation in integrated PET/MR systems, since the RF coil casing and contained electronic components are located in the field-of-view of the PET detector

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during independent PET and MRI data acquisition. This causes photon attenuation, which may introduce a bias in tracer activity quantification with PET .

While any RF coil for use in PET/MR in general shall be optimized for PET transparency , i.e. provide lowest possible photon attenuation, additional methods for attenuation correction (AC) are warranted to provide accurate tracer activity quantification with PET . In former stand-alone PET systems, AC was achieved by a transmission scan with ^{68}Ge rod sources that provided a spatial distribution of linear attenuation coefficients (LAC) across the patient tissues . In PET/computed tomography (CT) hybrid imaging, AC is performed by using the CT transmission data to calculate the attenuation in human tissues. In PET/MR, attenuation correction of patient tissues and hardware components such as RF coils has to be solved in different ways since transmission data either from ^{68}Ge rod sources or from CT is not available .

Several approaches for attenuation correction in PET/MR have been established. Patient tissues are attenuation corrected by using MR-based VIBE (volume interpolated breath hold examination) sequences with Dixon fat/water separation to generate μ maps of the patient tissues. This is the base for 4-compartment tissue segmentation into background air, fat, lung, and soft tissue . In addition, further MR sequences such as ultrashort TE (UTE) and zero echo time (ZTE) are used to provide cortical bone segmentation of the skull in PET/MR neuroimaging. Also atlas-based methods are used for bone segmentation in whole-body PET/MR applications .

For attenuation correction of hardware components such as RF coils or the patient table, CT-based μ maps of the hardware components are joined with the patient μ map before PET data reconstruction. The hardware component μ maps are based on CT transmission scans of the RF coils and provide the spatial distribution of attenuation factors in a 3D model of the respective hardware component . Due to the patient table position, the position of rigid RF coils (e.g. RF head coil, RF spine array coil) during a PET/MR examination is known and the template-based AC using the CT-derived μ map can be performed during the PET data reconstruction process .

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Among the wealth of MR imaging (MRI) and MR spectroscopy (MRS) methods, ^{13}C NMR is the method of choice for studying brain and cancer metabolism . Complementing structural proton (^1H) MR imaging, ^{13}C carbon MRS provides additional diagnostic criteria for tumour characterization and patient stratification . The combination of integrated PET/MR with ^{13}C MRS and MRI potentially offers simultaneous dual-modality metabolic imaging . Due to the inherently low natural abundance of endogenous ^{13}C , however, hyperpolarization techniques are needed to increase the ^{13}C signal in MRI and MRS to clinically usable levels . Hyperpolarization of ^{13}C is only recently available at selected sites. The combination of ^{13}C hyperpolarization and PET/MR hybrid imaging is of high potential interest in multi-modal, multi-parametric neuroimaging, but due to technical hurdles, this has not been fully realized in clinical applications yet . One missing link before clinical application on humans is the fact, that no radiofrequency (RF) head coil for $^{13}\text{C}/^1\text{H}$ MR signal reception for use in PET/MR is available today.

In this study, a new dual-tuned $^{13}\text{C}/^1\text{H}$ RF head coil was designed to be used in integrated PET/MR hybrid neuroimaging. For attenuation correction of the RF head coil, CT-based attenuation templates were acquired and implemented into the PET data reconstruction process. Systematic evaluation of the RF coil and its attenuation correction was then performed in phantom and patient examinations.

Materials and Methods

PET/MR hybrid system

All phantom and patient measurements were performed on an integrated 3-Tesla (3T) PET/MR whole-body hybrid system (Biograph mMR; Siemens Healthcare GmbH, Erlangen, Germany), which allows for simultaneous PET and MR imaging .

PET/MR dual-tuned head imaging RF coil

Basis for the development of a dual-tuned $^{13}\text{C}/^1\text{H}$ RF head coil for use in integrated PET/MR was a product 3T dual-tuned RF head coil for MR-only use (Rapid Biomedical GmbH, Rimpar, Germany). The MR-only RF coil underwent several design optimizations to reduce PET signal attenuation and, thus, to increase its PET transparency for use in PET/MR. Major changes in the RF coil design were the following. The 16-rung birdcage antenna length was increased from 209 mm to 265 mm. This maximizes the length of the birdcage part of the RF coil to match the length of the integrated PET detector (259 mm) (1) (6). Increasing the length of the birdcage resonator required a retuning of both ^1H and ^{13}C resonance frequencies. Due to quadrature polarization, the RF coil is optimized for RF signal transmission and reception for both nuclei. The RF coil is dual-tuned to the Larmor frequencies of ^1H 123 MHz and ^{13}C 31 MHz at 3 Tesla, serving as volume RF coil for the two MR nuclei. All electronic components and connecting cables were positioned outside of the FOV of the PET detector, since solder and copper are strong attenuating materials. Instead of using copper tubes with 0.5 mm wall thickness as for the 16 rungs of the MR-only birdcage, circuit boards with a 35 μm copper layer were used as resonators in the redesigned PET/MR RF head coil (Figure 1). The wall thickness of the RF coil casing was reduced to minimize the attenuation of the photons. The outer end-ring of the birdcage antenna, which is located near to the patient shoulders, was designed thinner than in the MR-only product version of the RF coil. The coil casing was designed to fit at a defined position on the patient table of the PET/MR system. This is a prerequisite for accurate attenuation correction applying CT-based 3D attenuation templates, which can only be achieved when the hardware μmap during the PET data reconstruction and the RF head coil on the patient table during PET data acquisition exactly match in their positions (refs 6, 12, 13, 14, 28).

The dual-tuned $^{13}\text{C}/^1\text{H}$ birdcage resonator is equipped with filters and decoupling mechanisms for enabling wideband alternating-phase low-power technique for zero-residual splitting (WALTZ) decoupling . Dual-tuned RF coils equipped with spin decoupling when using WALTZ

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sequences in principle enables increasing the signal-to-noise in NMR spectra while reducing the specific absorption rate (SAR) .

Phantom measurements

Phantom experiments were performed to evaluate functionality and accuracy of attenuation correction and to provide a means for systematic quantification of PET signal attenuation parameters. A cylindrically shaped phantom (volume 9480 ml) was used. The phantom is equipped with a handlebar for transport and can additionally be filled with radiotracer by ports on the top lid, thus providing a homogeneous large-volume emission source.

The PET/MR hybrid system provides a dedicated phantom holder for the cylindrical phantom. The phantom holder enables the phantom to be fixed facing the head end of the patient table. This mounting holds the phantom in a free-floating position several centimeters above the head end of the patient table in co-axial orientation to the table. The height of the phantom floating over the table can be chosen such that the RF head coil can be placed on the table and surrounding the phantom. Alternatively, the RF coil can be removed without repositioning the phantom. Thus, the phantom can be used as homogeneous and large-volume PET emission source, enabling difference measurements with and without the RF head coil in place.

To serve as PET emission source in this study, the phantom was filled with ^{18}F radiotracer for all following phantom scans with an initial activity concentration of 13 kBq/ml, resulting in a total activity of 124 MBq for the whole phantom. The phantom was centered with its longitudinal axis along the longitudinal axis of the RF coil (Figure 2A). The scan time for one measurement was 30 min. The rather high phantom activity and rather long acquisition time in the phantom experiments were chosen such that the difference measurements with and without RF coil in place provide sufficient PET signal statistics and homogeneity to accurately evaluate the attenuating effects of the RF coil hardware. PET data was recorded in list-mode.

Two phantom measurements were executed to obtain difference measurements (Figure 2). Both sets of experiments were time- and decay-corrected to compensate the different measurement starting points and for tracer decay over time.

Attenuation Correction

An established method for the attenuation correction of human tissues in PET/MR is based on MR measurements . Based on a Dixon-VIBE sequence, fat and water signal images are acquired to segment the patient tissue into four different classes: fat, soft tissue, lung and air. The MR parameters for the Dixon-VIBE sequence used in this study are as follows: image matrix 192 x 192; spatial in-plane resolution 2.6 x 2.6 mm²; slice thickness 3.12 mm; TR 3.6 ms, TE₁ 1.23 ms and TE₂ 2.46 ms; flip angle 10°, acquisition time 19 sec/bed position.

For attenuation and scatter correction of hardware components and RF coils in PET/MR, the method of using different CT-based attenuation templates has been established . During the phantom experiments in this study, the MR-based Dixon VIBE images of the phantom only serve as the basis for position detection of the phantom during the experiments. This position was then used as a marker to accurately co-register the CT-based μ map template for attenuation correction of the phantom during PET data reconstruction. Thus, the Dixon MR data was only used for position detection; the CT-data was used for attenuation correction of the phantom as described below. The MR-based data was not directly used for attenuation correction of the phantom since the phantom casing and other attenuating parts of the phantom do not provide MR signal and, thus, cannot be accurately corrected .

The CT imaging for the RF head coil (Figure 3A) and the phantom (Figure 3B) was performed on a dual source CT scanner (SOMATOM Definition Flash, Siemens Healthcare GmbH, Erlangen, Germany). The following parameters were used: tube voltage 140 keV, tube current 500 mA, matrix size of 512 x 512 pixels, voxel size 0.3 x 0.3 x 0.6 mm³ and a B30f-smooth convolution kernel. The CT data with an energy window level for the photons of 140 keV was converted to the PET energy level

of 511 keV with a bilinear conversion function .

This conversion from Hounsfield units (HU) to linear attenuation coefficients (LAC) and the following steps such as signal thresholding, artifact removal, and subsequent Gauss filtering were performed with MatLab 2013b (MATrix LABoratory, Mathworks, Massachusetts, USA) using a custom-written program. The following image processing steps were performed to exactly match the CT-based model with the fixed position of the RF head coil on the patient table. A signal threshold was applied to CT data to reduce CT streak artifacts around highly attenuating components. To match the high spatial resolution of the CT data with the lower spatial resolution of the PET detector (4.0 - 6.0 mm), a Gaussian filter was applied. For registration and matching of the μ map with the exact position of the RF coil on the patient table Vinci (Vinci Version 4.63, Max-Planck-Institute for Metabolism Research, Cologne, Germany) was used.

All post-processing and data reconstructions of the PET data along with the AC μ maps of the table, the cylindrical phantom and the custom μ map of the RF head coil were performed offline (Figure 4A) using a specialized vendor-provided software (e7 tools, Siemens Molecular Imaging, Knoxville, USA).

The PET data imaging matrix was $344 \times 344 \times 127$ with a voxel size of $2.08 \times 2.08 \times 2.03 \text{ mm}^3$. These parameters follow the standard 3-dimensional ordinary Poisson ordered-subsets expectation maximization (3D OP-OSEM) with 3 subsets and 21 iterations reconstruction parameters of the PET/MR hybrid system. For a full reconstruction of the PET data, the hardware AC map of the patient table is added. Furthermore, the CT-based 3D template model of the phantom is added (Figure 3B). The CT-based μ map of the RF head coil is added as a custom μ map ($\text{PET}_{\text{with coil}}$). The real position of the RF coil on the patient table is mechanically fixed. Hence, the μ map position during the PET data reconstruction process is linked with the known position of the RF coil on the patient table, which is

a precondition to obtain accurate results in attenuation correction.

The reconstruction of the second setup ($PET_{\text{without coil}}$) contains the CT-based μ map of the phantom and the automatically added patient table. Since the RF coil was removed in this case (Fig. 2B), its μ map is not used for the attenuation calculation (Figure 4B). The resulting μ maps of the two reconstructed data sets are then evaluated in terms of their difference (Equation 1), since any signal level changes should result from the presence of the local RF coil due to the otherwise unchanged setup.

$$\text{relative difference maps} = \frac{PET_{\text{with coil}} - PET_{\text{without coil}}}{PET_{\text{without coil}}}$$

Equation 1: Comparison of two PET measurements for pointing out the influence of photon attenuating hardware

Patient measurements

Seven patients were examined with the dual-tuned quadrature driven $^{13}\text{C}/^1\text{H}$ head coil. Six patients underwent a hybrid PET/MR ^1H examination, whereas patient #7 was scanned with the ^{13}C option. PET/MR hybrid imaging using ^1H MR and ^{18}F -FDG as PET-radiotracer was performed on six patients (4 male, 2 female; 68 years \pm 6 years; 168 cm \pm 16 cm; 71 kg \pm 11 kg; 285 MBq \pm 50 MBq; 3 h 16 min post injection \pm 46 min). The patient characteristics are provided in Table 1.

For simultaneous PET and MR imaging two sequences were acquired: first, a Dixon-VIBE sequence with subsequent tissue segmentation provides the soft tissue μ map for attenuation correction of the head region; second, a transversal T2-weighted TIRM dark fluid sequence (FLAIR) provides anatomical ^1H MR imaging with higher spatial resolution. The two MR sequences had following parameters Dixon-VIBE: FOV 500*500 mm², image matrix 192*192; spatial resolution 2.6 x 2.6 mm²; slice thickness 3.12 mm; TR 3.6 ms, TE1 1.23 ms and TE2 2.46 ms; flip angle 10°. The parameters of the FLAIR sequence were: TR 4480 ms, TE 60 ms, TA 45 sec, slice thickness 2.0 mm,

FOV 340*340 mm², matrix 256*256, FA 89°, acceleration factor 3 (GRAPPA).

To enable PET difference measurements, patient measurements were performed twice; each PET/MR measurement lasted 3 minutes. Accordingly, the first measurement was performed with the RF coil mounted at the head end of the patient table, surrounding the head of the patient. Subsequently, the RF coil was removed from the patient table and thus from the FOV of the PET detector to provide a second PET measurement without attenuation due to the RF coil. In this case, the built-in RF transmit/receive body coil of the PET/MR served as receiver coil for the MR sequences. For this second measurement, the head of the patient rested on a low attenuating pillow to achieve a similar position as in the first measurement when using the RF head coil. Thus, it was ensured that differences in the head positions between the two exams are minimized. Residual spatial differences (shifts or rotations) of the PET data sets were compensated by co registration using Vinci (Vinci Version 4.63, Max-Planck-Institute for Metabolism Research, Cologne, Germany). The comparison between the two setups was computed using a customized program, written in MATLAB to depict the relative difference, caused by the attenuating RF coil.

Attenuation correction in the first setting with the RF head coil in place was performed using a three-component AC map during PET data reconstruction (Figure 5A). For the second patient setting without RF head coil in place, the AC map of the RF head coil was removed from the PET data reconstruction (Figure 5B).

The matched and co-registered patient PET data sets were reconstructed applying the individual μ map, and the relative difference between the two PET data sets was calculated. The relative differences are a measure of the photon attenuation caused by the dual tuned RF head coil. Overall image quality of MR, PET, and PET/MR hybrid images was independently evaluated by two experienced readers (physicist and neurologist, each 20 years of MR and 8 years of PET/MR experience). In MR images, overall signal homogeneity, soft tissue contrast and image quality were subjectively rated. In PET images, signal homogeneity, and spatial detail of tracer distribution were

subjectively rated. All MR and PET images were additionally inspected for artifacts.

The dual-tuned quadrature driven $^{13}\text{C}/^1\text{H}$ RF head coil enables ^{13}C MR imaging and MR spectroscopy. Therefore, one additional patient (patient #7) was scanned with the ^{13}C option and a ^{13}C spectrum of the whole brain was acquired using a free-induction decay (FID) sequence. A hard rectangular $100\ \mu\text{s}$ excitation RF pulse was used to achieve a flip angle of 90° . There was a $100\ \mu\text{s}$ delay between excitation and acquisition, TR was chosen to be 1500 ms along with 206 averages, receiver bandwidth = 5000 Hz, two-step phase cycling, resolution = 2048 points, which resulted in an acquisition time of approximately 5 minutes. Dual-echo GRE shimming was used in order to optimize B_0 homogeneity and the obtained peak line-width .

Results

Phantom measurements

Figure 6 shows the relative PET difference maps of the cylindrical emission phantom in all three spatial directions. Differences between the two PET scans are visualized according to the difference map calculation (Equation 1).

Figure 6A depicts the relative difference between the PET measurement where the RF coil is placed in the PET FOV and the corresponding case without any RF coil (reference scan). The mean difference in relative activity is $+8.82\% \pm 0.38\%$, reflecting the overall attenuation caused by the RF head coil. The local differences across the phantom volume range from 1 % up to 25 %. The highest local difference is found in the lower region of the RF coil close to the patient table where the head-neck support cradle causes highest attenuation. The open coil space near the upper birdcage rungs causes less attenuation in the upper regions of the phantom volume.

Figure 6B shows the relative difference results of the two PET measurements with and without the RF coil in place, after applying the attenuation correction μ map of the RF coil. Thus, Figure 6B shows the attenuation correction influence of the CT-based μ map of the RF coil. The white, light red and light blue colors in the difference maps indicate a successful attenuation correction. Light red regions show a slight under-correction, light blue areas indicate a slight over-correction mostly caused by statistical noise.

Figure 6C shows the measured mean value of 75 ROIs in z-direction over 150 mm of the phantom (stack of green ROIs in coronal view). The non-attenuated-corrected ROI (red graph) varies around $+8.82 \% \pm 0.38 \%$. The AC ROI (blue graph) varies around $0.96 \% \pm 0.29, \%$ which demonstrates successful attenuation correction.

Patient measurements

Example MR, PET, and fused PET/MR images of three patients are shown in Figure 7. The PET images are attenuation corrected with the MR-based AC patient μ map, the CT-based AC of the systems patient table, and the custom-built AC for the RF head coil (Fig 7B). The fused PET/MR images show anatomical information from the MR images, along with the radiotracer distribution from the PET acquisition (Fig 7C). Overall image quality of MR, PET, and PET/MR hybrid images was rated good to excellent and of diagnostic quality. MR signal homogeneity and image contrast on FLAIR images was rated excellent, image quality was not hampered by noise, and no artifacts were detected. In PET images, signal homogeneity and spatial detail of tracer distribution were rated good and of diagnostic quality despite the fact that the PET/MR exams were started with a significant delay time (3h 16min post injection ± 46 min).

Figure 8 shows the PET images and difference maps of three representative patients. In these measurements, the average difference across all six patient PET measurements without RF coil (reference, Fig. 8A) and patient scans with non-corrected RF coil attenuation (Fig. 8B) of all six

patients was measured to $5.17 \% \pm 0.81 \%$ (MEAN \pm SD), which is also visually indicated in Fig. 8C by the light red color scale.

The difference maps in Figure 8D visually indicates that all relative differences are close to around zero percent (white), reflecting successful attenuation correction of the RF head coil. Measurements across the brain PET images of all six patients determined the mean differences to the reference scans after RF coil AC to $-0.87 \% \pm 0.65 \%$. Figure 8E shows the mean value of the relative differences measured in all brain volumes along the axial axis. The mean values in these ROIs in all patients show a $5.17 \% \pm 0.81 \%$ difference between PET data acquired without RF coil (reference) and with RF coil (without AC) as red curve. Following AC of the RF head coil, the mean difference values are $-0.87 \% \pm 0.65 \%$ (blue lines).

The ^{13}C spectrum of the non-selective brain scan as acquired in patient #7, demonstrates the functionality of the dual tuned RF head coil (C) providing a ^{13}C spectrum with high signal-to-noise and narrow peak line-width (Fig. 9).

Discussion

In this study a dual-tuned $^{13}\text{C}/^1\text{H}$ RF head coil for integrated PET/MR neuroimaging was developed, implemented into a PET/MR system and evaluated. The RF head coil was designed for optimized PET transparency. Attenuation correction of the RF head coil was performed by generating and applying a dedicated 3D CT-based template μmap . Attenuation correction was systematically evaluated in phantom experiments using a large-volume cylindrical emission phantom filled with 18-F-FDG radiotracer. The PET/MR imaging performance and PET attenuation correction were then evaluated in six patients. The functionality of the dual-tuned $^{13}\text{C}/^1\text{H}$ head coil was further demonstrated by generating ^1H anatomical MR images and ^{13}C MR spectroscopy of the brain for one patient.

The phantom measurements revealed that the RF head coil causes an overall attenuation of about $+8.8 \% \pm 0.38 \%$, reflecting the global attenuation caused by the RF head coil. The local differences across the phantom volume range from 1 % up to 25 %. The highest attenuation values were found in the lower region of RF coil close to the patient table where the head/neck support cradle attenuates photons the most. The open coil space near the upper birdcage rungs causes less attenuation in the upper regions of the phantom volume. The attenuation profile across the phantom volume as displayed on the resulting difference maps shows a rather homogeneous attenuation, reflecting the open birdcage design of the RF coil. Homogeneous and low-level attenuation of an RF coil can be seen as advantageous features and a precondition for subsequent attenuation correction . The overall attenuation caused by the RF head coil of 8.8% as measured in the phantom experiments can be considered as comparably low, when compared to the results of other studies. MacDonald et al. have determined the photon attenuation caused by a MR-only head RF coil to 20% . Tellmann et al. in another study have investigated the attenuation caused by a dedicated PET/MR head RF coil. They have found an attenuation bias in the range of 13-19% . The article by Paulus and Quick provides a comprehensive overview about numerous studies investigating the attenuation of several hardware components for use in PET/MR. No studies are available today specifically investigating dual-tuned RF head coils for PET/MR use.

Following attenuation correction by applying the CT-based AC template of the RF head coil, the phantom measurements revealed a measured difference across the phantom volume of $0.96 \% \pm 0.29 \%$, which is the measured difference between phantom measurement without RF coil and a second phantom measurement with RF coil and its attenuation correction. This rather low remaining quantification bias indicates successful attenuation correction of the RF head coil and a successful implementation of the RF coil on the PET/MR system and in the PET data reconstruction process .

The results of the patient measurements corroborate to a large extent the results of the phantom measurements. Anatomic MR images and PET images were all rated of high quality. Artifacts were neither observed in MR nor in PET images. Difference between a first set of PET measurements without RF coil and a second set of PET measurements with the RF coil in place showed a mean value of $+5.1\% \pm 0.81\%$ for all six patients when no attenuation correction was applied. Following attenuation correction with the 3D RF head coil attenuation correction template, the mean difference between both measurements was reduced to difference values around $-0.87\% \pm 0.65\%$. As in the phantom experiments, this reduction of bias in the difference measurements indicates successful AC of the RF head coil. It has to be noted, however, that the difference maps in Figure 8 show local fluctuations (red and blue areas) due to short PET acquisition times (3 minutes) and a rather long delay between tracer injection and PET data acquisition (3 h 16 min post injection ± 46 min). A slight over-correction of the PET images can also be based on scatter correction for the μ maps. Furthermore, two sequential patient measurements with and without the RF coil in place inevitably lead to a slight shift in the head position between the two measurements. For an accurate difference measurement, the two head scans need to be accurately coregistered.

The measurement of a non-selective ^{13}C MR spectrum in a patient serves as a general demonstration that the dual-tuned RF head coil is also able to acquire ^{13}C spectra with high signal-to-noise and sharp peak line-width. To unveil the full potential of PET/MR combined with ^1H and ^{13}C MRI/MRS in neuroimaging, however, further studies utilizing also ^{13}C hyperpolarization are needed. The natural abundance of ^{13}C is low; consequently, the SNR of ^{13}C signal in MRI and MRS is rather low as well. ^{13}C hyperpolarization can provide a strong boost of the ^{13}C signal. Thus, the full diagnostic potential for such a dual-tuned RF coil is provided in a setting where ^{13}C hyperpolarization and a PET/MR system are available. This combination would allow for ^1H high-resolution MRI and simultaneous PET/MR hybrid imaging in neuroimaging applications. In addition, the availability of hyperpolarized ^{13}C MRI and MRS could provide important information on ^{13}C tumour metabolism

adding important diagnostic information in neuro-oncologic applications.

Today, no dual-tuned RF head coil for PET/MR neuroimaging is available yet. In this context, the development of the dual-tuned RF head coil and the acquisition of a dedicated attenuation correction for this specific RF coil as presented in this study serves as an important step providing technical and methodological preconditions for multinuclear $^{13}\text{C}/^1\text{H}$ PET/MR hybrid neuroimaging.

Conclusion

A dual tuned $^{13}\text{C}/^1\text{H}$ RF head coil was designed and evaluated regarding its potential use in integrated PET/MR hybrid imaging. An according CT-based attenuation correction template of the RF head coil was generated. Attenuation correction was evaluated in phantom experiments and in a patient study on six patients and provided accurate results for PET quantification. In conclusion, the dual-tuned RF head coil was successfully integrated into PET/MR hybrid imaging and can now be used for advanced $^{13}\text{C}/^1\text{H}$ multinuclear hybrid neuroimaging in future studies.

Financial disclosure

Michael Sauer and Titus Lanz, Ph.D., are both employees of Rapid Biomedical GmbH (Rimpar, Germany). Radhouene Neji, Ph.D. is an employee of Siemens Healthcare Limited (Frimley, UK).

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Figure captions

Figure 1: Dual-tuned $^{13}\text{C}/^1\text{H}$ radiofrequency head coil that was designed for use in PET/MR hybrid imaging. (A). T1 transversal dark fluid (FLAIR) 1H MR image of a healthy volunteer in transversal orientation (B). T1 MPRAGE in sagittal orientation (C).

Figure 2: Phantom experiment to measure PET signal attenuation caused by the RF coil being placed in the field-of-view of the PET detector. Two measurements were performed: with RF head coil (A) and without RF coil placed around the active transmission phantom (B). Thus, difference maps can be acquired to determine the spatial distribution of PET signal attenuation caused by the RF coil across the homogeneous phantom volume. During the phantom experiments, the centers of the RF coil and phantom were centered to the middle of the PET detector.

Figure 3: A CT-template of the dual-tuned head coil (here as 3D rendering) shows that the most-attenuating electronic components are located in the wider casing to the left which is positioned outside of the PET detector (A) during PET/MR imaging. The CT-based attenuation template of the cylindrical water transmission phantom was applied in all phantom measurements for AC of the phantom (B).

Figure 4: Combined hardware μ maps for attenuation correction in the phantom experiments. CT-based μ map of the RF head coil combined with CT-based μ map of the water phantom and patient table (A). To provide a reference for difference measurements, a setup is measured where the RF coil is removed but the patient table and the phantom are attenuation corrected (B).

Figure 5: Attenuation correction μ maps reflecting two setups for the patient measurements. The complete μ map in (A) shows the experimental setting for patient measurements with the RF head coil in place. The μ map in (B) shows the setting for the reference measurement without the RF coil.

Figure 6: Relative difference maps comparing two phantom measurements with/without RF coil placed in the field-of-view of the PET detector during PET data acquisition (A). Local attenuation caused by the RF coil in the phantom is up to 25% (red areas in lower phantom region (A)). Applying CT-based attenuation correction of the RF coil results in a reduced quantification bias, thus reducing the differences to the PET measurement without RF coil to almost zero (B). The graph in (C) provides mean difference values in % for (A) and (B). The mean difference value was measured in a large region of interest (ROI, green circle in (A)) in 75 slices along the longitudinal direction of the phantom (z-direction). The mean attenuation value caused by the RF coil is 8.8 % (red line in (C)), which decreases to a mean value of 0.96% after applying attenuation correction for the RF coil (blue line in (C)).

Figure 7: Examples for PET/MR hybrid imaging in three representative patients. Transversal view of MR images (FLAIR) (A) acquired with the ^1H proton channel of the dual-tuned RF head coil. The MR images show good image quality and homogeneous signal distribution. Attenuation corrected PET-only images (B). Fused anatomical ^1H MR and functional ^{18}F -FDG PET images (C) showing excellent spatial coregistration.

Figure 8: Impact of hardware component attenuation correction on PET quantification in patients. The PET images of three patient cases demonstrate the relative PET quantification bias caused by the RF head coil. Column (A) shows the transversal PET images of the reference measurement without RF coil serving as reference for quantification. Column (B) shows the PET images acquired with the RF head coil and after attenuation correction with the custom μ map of the RF coil. Column (C) provides a relative difference map of the reference scan and the non-AC images, which shows a visual quantification bias around 10% (light red color) and local visual differences up to 30% (red color). Column (D) shows relative difference maps providing the difference between the references scan (PET images acquired without RF coil) and images acquired with the RF coil after attenuation correction. Averaged mean difference values across the transversal slices show a distribution around 0% (white, light red and light blue color). The graphs in Column (E) show the averaged mean values of a stack of transversal slices in z-direction (feet-to-head direction), where a large ROI has been placed in each slice. The quantification bias for the non-AC images is around 5.1 % (red lines), while the bias is reduced to around -0.8 % following attenuation correction of the RF head coil (blue lines).

Figure 9: PET image of patient #7 (A) whom was scanned with the ^{13}C option of the double-tuned RF head coil. The fused PET/MR (B) image of the patient combines functional information provided by PET and anatomical information provided by MR (T2 TSE). The ^{13}C spectrum of the non-selective brain scan, demonstrates the functionality of the dual tuned RF head coil (C) providing a ^{13}C spectrum with high signal-to-noise and sharp peak line-width.

Table 1: Patient characteristics.

Patient	Sex	Age	Size	Weight	BMI	^{18}F -FDG	time post
		[years]	[cm]	[kg]	[kg/m ²]	Activity [MBq]	injection [min]
#1	F	62	153	69	29,5	235	138
#2	F	75	168	60	21,3	284	144
#3	M	67	184	82	24,2	335	178
#4	M	64	180	79	24,4	299	137
#5	M	66	176	77	24,9	315	152
#6	M	75	181	77	23,5	278	231
#7	F	53	160	49	19,1	342	119









