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Simultaneous carotid PET/MR: Feasibility and Improvement of Magnetic Resonance-based Attenuation Correction

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Abstract

Purpose—Errors in quantification of carotid positron emission tomography (PET) in simultaneous PET/magnetic resonance (PET/MR) imaging when not incorporating bone in MR-based attenuation correction (MRAC) maps, and possible solutions, remain to be fully explored. In this study, we demonstrated techniques to improve carotid vascular PET/MR quantification by adding a bone tissue compartment to MRAC maps and deriving continuous Dixon-based MRAC maps.

Methods—We demonstrated the feasibility of applying ultrashort echo time-based bone segmentation and generation of continuous Dixon MRAC to improve PET quantification on five subjects. We examined four different MRAC maps: system standard PET/MR MRAC map (air, lung, fat, soft tissue) (MRAC_{PET/MR}), standard PET/MR MRAC map with bone (air, lung, fat, soft tissue, bone) (MRAC_{PET/MRUTE}), continuous Dixon-based (MRAC_{CD}) MRAC map (no bone) and continuous Dixon-based MRAC map with bone (MRAC_{CDUTE}). The same PET emission data was then reconstructed with each respective MRAC map and a CTAC map (PET_{PET/MR}, PET_{PET/MRUTE}, PET_{CD}, PET_{CDUTE}) to assess effects of the different attenuation maps on PET quantification in the carotid arteries and neighboring tissues.

Results—Quantitative comparison of MRAC attenuation values for each method compared to CTAC showed small differences in the carotid arteries with UTE-based segmentation of bone included and/or continuous Dixon MRAC; however, there was very good correlation for all methods in the voxel-by-voxel comparison. ROI-based analysis showed a similar trend in the carotid arteries with the lowest correlation to PET_{CTAC} being PET_{PETMR} and the highest correlation to PET_{CTAC} being PET_{CDUTE}.

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Disclaimer: None

Conclusions—We have demonstrated the feasibility of applying UTE-based segmentation and continuous Dixon MRAC maps to improve carotid PET/MR vascular quantification.

Keywords

PET/MR; attenuation correction; Dixon; ultrashort echo time; carotid arteries

Introduction

Positron emission tomography (PET) and magnetic resonance (MR) imaging of the vasculature such as the carotid arteries are increasingly used as imaging endpoints in clinical trials of atherosclerotic treatments.[1-2] Typically, the PET and MR acquisitions for such trials are performed on separate PET/computed tomography (CT) and MR scanners. With the emergence of simultaneous PET/MR scanners, the ability to perform these acquisitions in one imaging session and lower cumulative doses to the patient, by replacing CT with MR, for longitudinal studies is highly advantageous. However, possible errors in quantification of PET in simultaneous PET/MR imaging when not incorporating bone as a tissue segment in MR-based attenuation correction (MRAC) maps may be problematic due to their proximity to the cervical spine.[3–11] Ripa, et al. previously demonstrated very good correlation of 18F-Fluorodeoxyglucose (FDG) standardized uptake values (SUV) in the carotid arteries in a comparison between PET/MR and PET/CT.[11] However, as the authors state, it was difficult to determine quantification errors solely due to attenuation maps with no bone tissue segment from additional sources of error that may be introduced when comparing two separate scanners.[11] In addition to the lack of a bone tissue segment in all current MRAC maps outside the head, the system standard MRAC map on both simultaneous PET/MR scanners consists of one empirically assigned fat attenuation value for all voxels in the MRAC map that are determined as fat and one empirically assigned water attenuation value for all voxels in the MRAC map that are determined as soft tissue.[12] This is in contrast to a continuous scale CTAC map that provides attenuation information for each voxel directly from the CT acquisition in PET/CT imaging. Several methods combining ultrashort echo time (UTE) and/or Dixon MR acquisitions with image processing techniques have been demonstrated to add bone and continuous attenuation values into MRAC maps with success, however, their effects on quantitative carotid imaging have not been demonstrated to date. [13–19]

In this study, we demonstrated techniques to improve carotid vascular PET/MR quantification by adding a bone tissue compartment to MRAC maps and deriving continuous Dixon-based MRAC maps.

Methods

All subjects provided written informed consent to participate in the study and the protocol was approved by the local Institutional Review Board.

PET/MR Image Acquisition

Five subjects (median weight 77 kg; range: 75 – 109 kg) with either a diagnosis of cardiovascular disease, high cholesterol and/or hypertension were injected with a median dose of 369 MBq (range: 352 – 532 MBq) of 18F- FDG and imaged on the Siemens Biograph mMR (PET/MR, Siemens Healthcare, Erlangen, Germany) after a circulation time of 120 minutes (median 109 min; range: 95 – 139 min)[27] The carotid PET/MR exam consisted of the system standard Dixon MRAC acquisition immediately prior to the start of the PET acquisition. The Dixon MR acquisition provides fat and water separation to assist in the generation of a 4-segment MRAC map ($\text{MRAC}_{\text{PETMR}}$) (air, 0.0000 cm^{-1} , lung 0.0224 cm^{-1} , fat 0.0854 cm^{-1} , soft tissue 0.1000 cm^{-1}), as is standard for attenuation correction on the current clinically approved PET/MR scanner software (Siemens Biograph mMR acquisition software VB20, Siemens Healthcare, Knoxville, TN, USA).[12, 20] Immediately after the Dixon MRAC acquisition, a PET acquisition was performed for 8 min and 1 bed position centered over the left and right carotid bifurcation of each subject. Simultaneously, during the PET acquisition, we acquired a time-of-flight (TOF) MR angiography (Acquisition time 13:29 min, TR 21 ms, TE 3.69 ms, FA 25°, FOV phase 100%, FOV 220 mm, slice thickness 1.00 mm, slabs 4, slice/slab 60, voxel size $0.9 \text{ mm} \times 0.9 \text{ mm} \times 1.0 \text{ mm}$) to provide characterization of the carotid arteries and for use in tracing the carotid ROIs. No dedicated MR coils were present during the PET acquisition. Immediately after the simultaneous PET and TOF MR angiography acquisition were completed, without moving the subjects, we placed a dedicated carotid coil (Machnet B.V., Roden, The Netherlands) on the patient to provide increased signal to noise during the UTE MR image acquisition (acquisition time 100 s, TR 11.94 ms, TE1 0.07 ms, TE2 2.46 ms, FA 10°, 1 slab, slice/slab 192, LR FOV 300 mm, AP FOV 300 mm, voxel size 1.6 mm isotropic). The dedicated carotid coil was only placed on the patient after the PET acquisition to avoid attenuation from the coil during PET acquisition.

CT-based attenuation correction map generation

Immediately after the PET/MR exam, to provide CTAC of the PET emission data acquired on the PET/MR scanner, each patient also had a low-dose CT acquisition (estimated maximum effective dose: 2.1mSv, voltage 120kVp, current 60mA/slice, voxel size $0.97 \text{ mm} \times 0.97 \text{ mm} \times 1.0 \text{ mm}$, image size $512 \times 512 \times 45$) on the Biograph mCT (PET/CT, Siemens Healthcare, Knoxville, TN, USA). Each CT image was co-registered to the respective Dixon water image of each patient from the PET/MR acquisition. To facilitate co-registration of MR and CT, subjects were placed in an identical head cushion during both acquisitions. Co-registration was performed in SPM8 (SPM, University College London, London, UK) using normalized mutual information estimation. Spatial accuracy of the co-registration was assessed by calculating the distances between 3 points (1 spine, and 2 carotids), in 5 axial slices, visible in both the Dixon water and CT images for each patient. The registered CT images were transformed to 511 keV attenuation coefficients using a bi-linear model implemented in MATLAB (Mathworks, Natick, MA).[21] The transformed CT images were then smoothed with a 4 mm Gaussian to match the PET resolution and re-sliced into the PET/MR MRAC image space.

Continuous Dixon Image Generation

In order to compare the effects of applying two empirically assigned tissue compartments each with a single value versus continuous attenuation coefficients more similar to a CT image, we processed the water and fat magnitude images using method similar to a previous study done in brain PET/MR. We calculated the water-fat fraction (eq. 1), as previously described by Berker, et al.[15]

$$r = \frac{M_w - M_f}{M_w + M_f} \quad (1)$$

Where M_w is the magnitude of the Dixon water MR image and M_f is the magnitude of Dixon fat MR image. The subsequent assignment of continuous attenuation values ranging from pure fat ($r = -1$) to pure water ($r = 1$) was performed by assigning the empirical attenuation values for pure fat (μ_{fat}) and pure water (μ_{water}) from the PET/MR scanner (fat 0.0854cm^{-1} , water 0.1000cm^{-1}) (eq. 2).[12, 15]

$$\mu_r = \frac{1+r}{2} * \mu_{water} + \frac{1-r}{2} * \mu_{fat} \quad (2)$$

The generated continuous Dixon MRAC maps (MRAC_{CD}) were then smoothed with a 4 mm Gaussian to match the PET resolution and re-sliced into the PET/MR image space.

Bone segmentation from Ultrashort Echo Time Acquisitions

Post-processing of the UTE images was done offline to provide our own UTE-based MRAC maps ($\text{MRAC}_{\text{PETMRUTE}}$ and $\text{MRAC}_{CD\text{UTE}}$) as this sequence is not currently optimized by the vendor to provide bone segmentation outside of the skull. The bone segmentation employed in this study was similar to the previous method of Keereman, et al.[13] Briefly, the $R2^*$ values within the neck were calculated with the calculation in equation 3 with UTE echo times of $\text{TE}_1 = 0.07$ ms and $\text{TE}_2 = 2.46$ ms.

$$R2^* = \frac{\ln(I_1) - \ln(I_2)}{\text{TE}_2 - \text{TE}_1} \quad (3)$$

After generation of the $R2^*$ maps, bone was segmented using a threshold of 0.35 ms^{-1} with morphological filtering that included neighboring voxels within a range of $0.15 - 0.50 \text{ ms}^{-1}$ assigned as bone. The segmentation algorithm was restricted to 20 slices above and below the carotid bifurcation and radially out from the center of the spine in an axial slice to prevent inclusion of areas distant from the spine such as the trachea or air in the sinus that also have high $R2^*$ values. All pixels that were determined to be bone were empirically assigned a bone attenuation value of 0.130 cm^{-1} . [13] The generated UTE-based bone segmentation was then inserted into both the 4-compartment system standard MRAC map ($\text{MRAC}_{\text{PETMRUTE}}$) and the continuous Dixon MRAC map ($\text{MRAC}_{CD\text{UTE}}$). Each final MRAC image was then smoothed with a 4 mm Gaussian to match the PET resolution and re-sliced into the PET/MR image space.

Accuracy of bone segmentation was determined by thresholding the bone compartment of the co-registered CT images at 300 Hounsfield units. The segmented CT images were considered the true bone segmentation for reference. Subsequently, all voxels thresholded as bone from the UTE-based segmentation were compared to the segmented CT images. To assess the accuracy of the segmentation we found the proportion of pixels in the UTE segmented as bone, using the CTAC as the reference. If the bone was segmented in the CT but not segmented by UTE this was reported as less than 100% segmentation and when voxels segmented in the UTE but were not determined as bone in the thresholded CT bone this was reported as greater than 100% segmentation, to report the possibility of under- and overestimation of bone by the UTE-based segmentation technique.

PET reconstructions

The acquired PET/MR emission data for each patient was reconstructed with five separate attenuation maps. All PET reconstructions were performed using vendor provided OSEM algorithm with 3 iterations and 21 subsets and corrections for normalization, dead time, decay, scatter, random coincidences, sensitivity and attenuation (e7tools, VB20, Siemens Healthcare, Knoxville, TN, USA). For the following comparisons, we reconstructed the original PET/MR data with the system standard 4-segment Dixon map ($MRAC_{PETMR}$), a CTAC map, continuous Dixon attenuation value map ($MRAC_{CD}$), standard mMR attenuation map with UTE bone segmentation added ($MRAC_{PETMRUTE}$) and continuous Dixon attenuation value map with UTE bone segmentation added ($MRAC_{CDUTE}$).

Image Analysis

Voxel-based analyses were performed using an in-house program written in MATLAB to calculate Pearson's correlations and Bland-Altman plots of all voxels within the PET images, using the PET_{CTAC} as reference. Voxel-based analyses were thresholded above 0.01 SUV to remove voxels dominated by noise. Region-based analyses were performed using an in-house program in MATLAB and Osirix (Osirix Imaging software, Pixmeo, Geneva, Switzerland) to determine differences in SUV within specific tissue regions. Six regions-of-interest (ROIs) were defined: left and right carotid arteries, left and right jugular veins, spine and soft tissue (muscle). ROIs were traced manually in the axial view of the TOF MR images for each ROI for 20 axial slices starting at the carotid bifurcation and descending in the caudal direction (Fig 1). These ROIs were then superimposed on all PET reconstructions to measure local quantitative differences. SUV mean, SUV maximum and Target-to-Background Ratios (TBR) (target is carotid; background is jugular vein) were calculated for each ROI. Pearson's Correlation and Bland-Altman plots were used to compare SUV mean, SUV maximum and TBR within each ROI of each patient. Statistical analyses to compare the results of each method were performed using SPSS software (SPSS, Chicago, IL). Pearson's correlation and paired t-test were evaluated to identify correlation and significant differences for each comparison. Statistical significance was considered for $p < 0.05$.

Results

Mean co-registration error of the CT to MR for all points was 3.7 mm (left carotid 4.1mm, right carotid 3.7mm and spine 3.2mm). Bone segmentation from the UTE MR acquisition

accurately segmented bone in 92.3% (43.6 – 153%) of voxels as compared to bone segmentation of the CT image. Qualitative comparison of MRAC maps showed obvious differences between CTAC maps and MRAC maps that did not include bone in attenuation maps (MRAC_{PETMR}, MRAC_{CD}) (Fig 2) while MRAC maps with bone included showed small differences in attenuation values in the bone compartments (MRAC_{PETMRUTE}, MRAC_{CDUTE}) (Fig 2). Quantitative comparison of MRAC attenuation values for each method compared to CTAC showed small differences in those with UTE-based segmentation of bone included (MRAC_{PETMRUTE}, MRAC_{CDUTE}) and larger differences when bone segmentation was not included (MRAC_{PETMR}, MRAC_{CD}), however, there was very good correlation for all methods in the voxel-by-voxel comparison (Table 1). The respective PET acquisitions showed small qualitative differences when bone compartments were included in the attenuation maps (PET_{PETMRUTE}, PET_{CDUTE}) compared to PET_{CTAC} and larger qualitative differences visible within the spine in the PET_{PETMR}, and PET_{CD} images compared to PET_{CTAC} (Fig 2 & Fig 3). Quantitative voxel-by-voxel comparisons for all MRAC PET reconstructions compared showed high correlation to PET_{CTAC} but the lowest correlation and the highest mean difference when comparing PET_{PETMR} to PET_{CTAC} ($y=0.85x+0.03$, $R^2=0.98$, mean difference $\pm$ SD = -0.10 ± 0.38 SUV, $p<0.001$) and the highest correlation and the lowest mean difference when comparing PET_{CDUTE} to PET_{CTAC} ($y=0.88x+0.05$, $R^2=0.97$, mean difference $\pm$ SD = -0.04 ± 0.37 SUV, $p<0.001$) (Fig 4, Fig 5, Fig 6). ROI-based analysis showed a similar trend as the voxel-by-voxel comparisons with the lowest correlation being PET_{PETMR} compared to PET_{CTAC}, ($y=0.88x$, $R^2=0.84$, $p<0.001$) and the highest correlation being PET_{CDUTE} to PET_{CTAC} ($y=0.97x$, $R^2=0.90$, $p<0.001$) (Fig 4, Fig 5, Fig 6) (Table 2). Median and range percent differences in ROI analyses were highest overall for the system standard PET_{PETMR} as compared to PET_{CTAC} (Table 3). ROI analyses showed smallest differences in the carotids with either PET_{CD} or PET_{CDUTE} and in the spine with PET_{CDUTE} (Table 3). When using TBR in the left or right carotid as a quantitative PET measure percent differences of all 4 MRAC methods were 4.3% overestimation compared to PET_{CTAC} (Table 4).

Discussion

This study compared the qualitative and quantitative effects of five different attenuation correction maps on the same PET emission data from a simultaneous PET/MR scanner. Mean co-registration of the CT image into the PET/MR image space was 3.7mm, which is less than the resolution of the PET scanner and thus should not have a large effect on quantitative results. Segmentation of bone using the UTE-based method resulted in an accuracy of 92.3% (43.6 – 153%) when compared to segmentation of bone from the CT image. Keereman, et al, using the same method in the skull achieved an accuracy of 85-95% of bone tissue correctly. [13] Berker, et al, achieved an accuracy of 70% while segmenting bone using their UTE triple echo sequence and segmentation methods. [15] Despite limiting our bone segmentation to slices directly below and above the carotid bifurcation we did not see quantification errors in the reconstructed PET images, from lack of bone in the chest or skull regions of MRAC maps, propagate to the carotids (Fig 2B & 3) (Table 2 & 3). We had a 30% lower segmentation threshold for UTE-based segmentation than Keereman, et al. [0.35ms^{-1} and range (0.15ms^{-1} to 0.55ms^{-1}) versus 0.50ms^{-1}]. [13] These threshold

differences could be explained due to the differences in echo times between our UTE sequence and the previous method, predicting $R2^*$ values of approximately 30% less.[13]

The largest gains in accuracy compared to PET_{CTAC} were seen when adding both the bone compartment and the continuous Dixon to MRAC maps ($MRAC_{CDUTE}$) in the spine ROI. We were able limit quantitative errors (median percent difference) in the spine ROI from -24.2% (range: -34.6% - 0%) to 2.1% (-8.3% - 15.5%), within the range of typical noise deviations resulting from statistical PET reconstructions, similar to previous studies.[13–16] It is clear that the addition of bone into the MRAC map improves quantification within bone structures; however, as seen in our quantitative analysis (Table 2, Table 3) further quantitative improvements (over PET_{CD}) are not extended to the vasculature (carotids and jugular veins) when adding bone to PET_{CD} (PET_{CDUTE}). This is in agreement with our previous work on a sequential PET/MR scanner for both clinical (carotid) and preclinical (rabbit aorta) studies assessing the accuracy of MRAC versus CTAC in the same PET emission data.[22–23] These studies demonstrate that quantification errors in MRAC from the lack of bone are limited to bone structures and areas immediately adjacent but do not extend to the carotid arteries or aorta.[22–23] Median percent difference errors using $MRAC_{CD}$ compared to CTAC improved underestimation of quantitative values for all vascular measurements (left carotid, -16.1% to -4.3%; left jugular, -12.1% to -3.2%; right carotid, -21.2% to -10.8%; right jugular, -20.8% to -12.8%), again to within the range of typical noise deviations from PET reconstruction, providing acceptable MRAC for carotid PET quantification. This may prove most beneficial clinically because the application of $MRAC_{CD}$ is relatively straightforward and does not require advanced MR acquisition or segmentation techniques to include bone in the MRAC map.

Previously demonstrated UTE methods have been limited to use in the brain where the detection of bone structures can be more straightforward versus detection and segmentation in the rest of the body.[13–15] We are the first to demonstrate the feasibility of accurately segmenting bone structures without atlas- or machine learning-based methods within a limited FOV in the neck.[13–19] Despite its possible utility in segmenting bone, a limitation of the UTE acquisition may be the limited acquisition field-of-view (FOV) due to sensitivity to distortions from field inhomogeneities at the outer edges of the MR acquisition FOV that may lead to errors in imaging and segmenting bone.[16, 24] Thus, it has been theorized that the use of UTE to assist in detecting bone outside the skull may be limited without atlas- and/or machine learning-based approaches to assist segmentation.[24] One previous study using machine learning-based methods did demonstrate bone segmentation and incorporation into the MRAC map in the pelvis.[16] In our case, the neck may be ideally suited for UTE-based MRAC maps without such methods due to the narrow FOV needed to image the width of the neck and spine; however, given our wide range of segmentation accuracy, a more advanced segmentation technique may improve these preliminary results. [16] In addition the presence of dedicated MR coils (carotid coil) for high resolution MR imaging during PET/MR exams will cause attenuation and degradation of PET images.[25] We have demonstrated the detection and subsequent attenuation correction of the dedicated carotid coil used in this study using the same UTE MR acquisition.[25] Thus, it may be possible to detect specific carotid MR coils and provide bone MRAC within the same UTE

acquisition sequence in carotid PET/MR imaging protocols but remains to be fully explored in clinical studies.

We chose to image the patients at 2 hours post-injection for the carotids based on previous vascular PET reproducibility studies in the carotid arteries which demonstrated TBR to be a highly reproducible measure of plaque inflammation at this time point.[26] Interestingly, when using TBR as a quantitative measure of PET uptake in the carotids, the cancellation of the blood pool by using each respective jugular vein resulted in less than 4.3% error in quantification irrespective of all AC methods (Table 4). TBR has been used in atherosclerotic plaque treatment clinical trials as a reproducible quantitative measure and would appear to be a candidate for quantitative measures in trials with PET/MR, as well.[1, 2, 26, 27] We limited all MR sequences used for image processing in the current study to sequences that are typically acquired simultaneous to carotid PET acquisition (Dixon and TOF MR Angiography) with the exception of the UTE MR acquisition. The addition of a UTE acquisition (100 seconds) is not a great hindrance to add to a carotid PET/MR imaging protocol. While determining the reproducibility of the techniques used here was not possible with our cohort of five patients the applicability of these methods, particularly, $MRAC_{CD}$ are in agreement with previous studies and could be easily implemented in a clinical setting to more accurately quantify inflammation in the carotid arteries using simultaneous PET/MR. There are several limitations for the current study, including the lack of pathology seen in this cohort. Despite this, we would hypothesize that plaque attenuation values may be characterized by the continuous Dixon method and more accurately than a system standard fat and water only separation. The continuous Dixon method may theoretically allow more accurate characterization of plaque AC between different plaques if they contain higher or lower water/fat fractions than surrounding voxels. Further characterization of the attenuation effects of plaques using these methods will need to be explored further in the future. In addition, the echo times of the UTE acquisition were optimized by the vendor for skull segmentation. Optimization of the echo times in the UTE acquisition for segmentation outside the skull may provide better raw $R2^*$ data for segmentation algorithms. Finally, the low number of subjects in this preliminary study limits warrants further study large clinical populations, with presentable disease on imaging. Larger cohort studies assessing continuous Dixon MRAC versus CTAC are needed to further verify the accuracy of carotid PET/MR as a possible endpoint in clinical trial imaging endpoint yet carotid PET/MR remains an attractive option due to the simultaneous acquisition and reduction in dose over longitudinal studies.

Conclusion

We have compared the PET/MR system standard $MRAC_{PETMR}$ maps versus CTAC maps in carotid PET/MR vascular imaging and demonstrated the feasibility of applying UTE-based segmentation of bone ($MRAC_{PETMRUTE}$, $MRAC_{CDUTE}$) and a continuous fat-water fraction from Dixon images ($MRAC_{CD}$) to create improved MRAC maps and more accurate carotid PET/MR vascular quantification.

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Figure 1.

Representative example of the six regions-of interest (ROIs) that were defined: left (red) and right (orange) carotid arteries, left (pink) and right (yellow) jugular veins, spine (blue) and soft tissue (muscle) (brown). ROIs were traced manually in the axial view of the time-of-flight MR images for each ROI for 20 axial slices starting at the carotid bifurcation and descending in the caudal direction.

Figure 2. Sagittal and axial comparisons of CT- and MR-based attenuation maps (A) and their respective PET reconstructions (B).

Figure 3. Percent difference error map showing regional differences in SUV between PET/MR emission data reconstructed with the respective MRAC and CTAC maps

Figure 4.

A Voxel-by-voxel Pearson's correlation plots for all voxels in all patients.

B Voxel-by-voxel Bland-Altman plots for all voxels in all patients. The mean (yellow dashed line) and the upper and lower limits (red dotted lines) are calculated according to a Bland-Altman analysis [28]. Upper limit is mean + 1.96*standard deviation and lower limit is mean – 1.96*standard deviation.

Figure 5.
Pearson's correlation plots for all ROIs in all patients.

Figure 6.

Bland-Altman plots for all ROIs in all patients. The mean (yellow dashed line) and the upper and lower limits (red dotted lines) are calculated according to a Bland-Altman analysis [28]. Upper limit is mean + 1.96*standard deviation and lower limit is mean – 1.96*standard deviation.

Table 1
Summary of the mean attenuation coefficients (cm⁻¹) within each ROI analyzed and for each of the five attenuation maps generated

Attenuation map	L Carotid	L Jugular	R Carotid	R Jugular	Muscle	Spine
CTAC	0.098	0.097	0.097	0.097	0.098	0.110
MRAC _{PETMR}	0.100	0.100	0.100	0.100	0.100	0.100
MRAC _{CD}	0.097	0.097	0.096	0.097	0.098	0.096
MRAC _{PETMRUTE}	0.100	0.100	0.100	0.100	0.100	0.110
MRAC _{CDUTE}	0.098	0.097	0.096	0.097	0.098	0.110

Table 2

Summary of the mean, minimum, and maximum SUV within each ROI analyzed for each of the five PET comparisons. SUV mean $\pm$ standard deviation (SUV min, SUV max).

SUV mean (min, max)	L Carotid	L Jugular	R Carotid	R Jugular	Muscle	Spine
PET_{CTAC}	1.32 $\pm$ 0.48 (0.89, 1.85)	0.93 $\pm$ 0.19 (0.69, 1.27)	1.22 $\pm$ 0.40 (0.83, 1.79)	0.90 $\pm$ 0.22 (0.66, 1.27)	0.64 $\pm$ 0.10 (0.42, 0.92)	1.23 $\pm$ 0.21 (0.59, 2.64)
PET_{PETMR}	1.20 $\pm$ 0.49 (0.80, 1.69)	0.84 $\pm$ 0.22 (0.62, 1.16)	1.08 $\pm$ 0.42 (0.71, 1.61)	0.77 $\pm$ 0.22 (0.54, 1.11)	0.60 $\pm$ 0.10 (0.41, 0.86)	1.03 $\pm$ 0.11 (0.51, 2.36)
PET_{CD}	1.28 $\pm$ 0.46 (0.86, 1.79)	0.90 $\pm$ 0.20 (0.67, 1.23)	1.15 $\pm$ 0.38 (0.77, 1.70)	0.84 $\pm$ 0.20 (0.61, 1.19)	0.62 $\pm$ 0.09 (0.42, 0.89)	1.07 $\pm$ 0.13 (0.52, 2.49)
PET_{PETMRUTE}	1.20 $\pm$ 0.50 (0.79, 1.69)	0.83 $\pm$ 0.21 (0.61, 1.15)	1.07 $\pm$ 0.43 (0.69, 1.60)	0.75 $\pm$ 0.22 (0.51, 1.09)	0.60 $\pm$ 0.09 (0.40, 0.89)	1.16 $\pm$ 0.14 (0.56, 2.51)
PET_{CDUTE}	1.29 $\pm$ 0.47 (0.87, 1.82)	0.90 $\pm$ 0.21 (0.66, 1.24)	1.16 $\pm$ 0.39 (0.77, 1.72)	0.84 $\pm$ 0.20 (0.61, 1.19)	0.63 $\pm$ 0.09 (0.43, 0.93)	1.25 $\pm$ 0.18 (0.59, 2.72)

Table 3

Summary of the median and range of percent differences calculated from SUV mean within each ROI analyzed, for each of the five comparisons.

	L Carotid	L Jugular	R Carotid	R Jugular	Muscle	Spine
PET _{PETMR}	-16.1 (-27.3 – 0)	-12.1 (-16.0 – 0.2)	-21.2 (-31.5 – 2.4)	-20.3 (-23.9 – 0.9)	-2.3 (-14.6 – 7.5)	-24.2 (-34.6 – 0)
PET _{CD}	-4.3 (-15.7 – 11.9)	-3.2 (-10.8 – 8.0)	-10.8 (-21.1 – 13.1)	-12.8 (-13.6 – 8.3)	-3.9 (-8.6 – 8.6)	-21.8 (-28.2 – -2.6)
PET _{PETMRUTE}	-18.3 (-24.3 – -0.1)	-14.5 (-19.6 – -0.9)	-21.0 (-30.7 – -0.8)	-20.5 (-28 – -1.3)	-2.0 (-11.8 – 6.9)	-10.3 (-17.8 – 4.8)
PET _{CDUTE}	-6.0 (-14.1 – 16.2)	-5.1 (-11.6 – 9.3)	-12.5 (-21.3 – 14.5)	-13.5 (-14.9 – 8.6)	-3.6 (-6.6 – 9.7)	2.1 (-8.3 – 15.5)

* PET_{CTTAC} as reference

Table 4

Summary of the target-to-background ratios (TBR) in terms of SUV mean measured within each carotid and its respective jugular vein for each of the five comparisons. TBR_{mean} (percent difference compared to PET_{CTAC})

	PET_{CTAC}	PET_{PETMR}	PET_{CD}	$PET_{PETMRUTE}$	PET_{CDUTE}
LC TBR (% diff)	1.42	1.43 (0.1%)	1.44 (1.4%)	1.44 (1.4%)	1.46 (2.8%)
RC TBR (% diff)	1.38	1.42 (2.9%)	1.40 (1.4%)	1.44 (4.3%)	1.41 (2.2%)