Comparison of quantitative \(^{11}\text{C}\text{PE2I}\) brain PET studies between an integrated PET/MR and a stand-alone PET system

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ABSTRACT

PET/MR systems demanded great efforts for accurate attenuation correction (AC) but differences in technology, geometry and hardware attenuation may also affect quantitative results. Dedicated PET systems using transmission-based AC are regarded as the gold standard for quantitative brain PET. The study aim was to investigate the agreement between quantitative PET outcomes from a PET/MR scanner against a stand-alone PET system.

Nine patients with Parkinsonism underwent two 80-min dynamic PET scans with the dopamine transporter ligand \(^{11}\text{C}\text{PE2I}\). Images were reconstructed with resolution-matched settings using \(^{68}\text{Ge}\)-transmission (stand-alone PET), and zero-echo-time MR (PET/MR) scans for AC. Non-displaceable binding potential (BP\(_{\text{ND}}\)) and relative delivery (R\(_1\)) were evaluated using volumes of interest and voxel-wise analysis.

Correlations between systems were high (\(r \geq 0.85\)) for both quantitative outcome parameters in all brain regions. Striatal BP\(_{\text{ND}}\) was significantly lower on PET/MR than on stand-alone PET (-7%). R\(_1\) was significantly overestimated in posterior cortical regions (9%) and underestimated in striatal (-9%) and limbic areas (-6%). The voxel-wise evaluation revealed that the MR-safe headphones caused a negative bias in both parametric BP\(_{\text{ND}}\) and R\(_1\) images. Additionally, a significant positive bias of R\(_1\) was found in the auditory cortex, most likely due to the acoustic background noise during MR imaging.

The relative bias of the quantitative \(^{11}\text{C}\text{PE2I}\) PET data acquired from a SIGNA PET/MR system was in the same order as the expected test–retest reproducibility of \(^{11}\text{C}\text{PE2I}\) BP\(_{\text{ND}}\) and R\(_1\), compared to a stand-alone ECAT PET scanner. MR headphones and background noise are potential sources of error in functional PET/MR studies.

1. Introduction

Hybrid positron emission tomography/magnetic resonance (PET/MR) systems have opened up many new neuroimaging opportunities, potentially providing a great tool for a broad range of research and clinical applications in neurology [1,2]. Initially, clinical neuroimaging applications for PET/MR have been hampered partially due to the lack of a widely accepted subject-specific attenuation correction (AC) method for the PET data [3]. The main reason is that MRI provides no direct information on electron density that can be used for photon AC [4,5]. In PET/computed tomography (CT) systems, a broadly accepted approach is a low-dose CT scan (CT-AC) [6], where the CT images are converted from Hounsfield units to linear attenuation coefficients at 511 keV [7]. Stand-alone PET systems often use \(^{68}\text{Ge}/^{68}\text{Ga}\) transmission sources for AC, which is generally regarded as the gold standard for quantitative brain PET imaging as it measures 511 keV photon attenuation directly [8,9].

In the past decade, extensive research has been performed to develop
and improve novel MR-based AC (MRAC) methods for brain PET/MR imaging [10–14]. Both Atlas- and direct imaging-based MRACs are currently commercially available in PET/MR systems. A major advantage of Atlas-based MRAC approaches is the use of continuous linear attenuation coefficients (LACs), but these methods are often time-consuming due to image registration to a database and cannot adjust for subject variability. Direct imaging MRAC methods based on image segmentation (into bone, soft tissue, and air cavities) are fast, have no need for image registration, and account for inter-subject variability. Disadvantages are the use of discrete LACs and sensitivity to artifacts. In this category, ultra-short echo time (UTE) [15], and zero-echo time (ZTE) [16–18] are favoured as they adequately detect and segment the skull bone. Consequently, these techniques allow adding subject-specific bone morphology to MRAC maps. In previous work, we demonstrated ZTE-MRAC being one of the best-performing MRAC methods compared to 68Ge-AC as the gold standard [19,20], which was in good agreement with other evaluations of MRAC methods [21,22]. Today ZTE MRAC is commercially available on specific PET/MR systems.

Apart from the choice of MRAC method, the different scanner technology of an integrated PET/MR, the scanner geometry, and hardware attenuation may affect the PET data. For example, disparities in the scanner’s geometry and existing attenuation material will have an impact on scatter fractions. Further, in contrast to a dedicated PET scanner, an integrated PET/ MR system has attenuating material between the patient and the PET detectors, such as the body coil in the scanner bed and the head coils. Although all the scanner bed and MR coils are usually addressed in the vendor provided MRAC maps, there may still be artefacts. Finally, small but measurable discrepancies between the PET performance with MR-idle and simultaneous scanning have been reported [23]. The aim of the presented work was to assess the agreement between quantitative outcome measures of a [11C]PE2I brain scan acquired on a PET/MR system and on a stand-alone PET scanner, both at VOI and voxel-level. The outcome of this study was expected to induce further development of various clinical applications on a PET/MR system. In addition, this study would provide knowledge how historical quantitative outcomes acquired on a stand-alone PET scanner relate to the results obtained on a PET/MR system which is relevant for retrospective studies.

## 2. Methods

### 2.1. Description PET imaging systems

PET images were acquired on both an ECAT Exact HR+ stand-alone PET scanner (ECAT; Siemens/CTI, Knoxville, Tennessee) and a SIGNA PET/MR (SIGNA; GE Healthcare, Waukesha, Wisconsin). The ECAT Exact HR+ scanner is a BGO scanner using three rotating 68Ge transmission line sources for attenuation correction. The SIGNA PET/MR consists of a 3T MRI and a digital time-of-flight capable PET system with lutetium-yttrium-oxyorthosilicate (LYSO) crystals coupled to silicon photomultipliers. The main physical characteristics of the ECAT and SIGNA scanners are summarized in Table 1.

### 2.2. Study design

This was a retrospective study and the participants, recruitment and inclusion criteria have been previously described [19]. Briefly, the data comprised nine patients with Parkinsonism (mean age 72 years, range 49–82; 5 females) and none had significant cerebellar atrophy. Each patient underwent first a dynamic [11C]PE2I brain PET scan on an ECAT scanner followed by a second scan on a SIGNA scanner within less than six months (mean: 73 days, range 12–147). The tracer [11C]PE2I binds highly selectively and specifically to central dopamine transporters (DAT) but also allows to study relative cerebral blood flow. The Regional Board of Medical Ethics in Uppsala and the Radiation Ethics Committee of Uppsala University Hospital approved the study, and all patients gave their written consent.

### 2.3. Scan protocol

An 80 min dynamic PET data acquisition consisting of 22 frames (4 × 60 s, 2 × 120 s, 4 × 180 s, 12 × 300 s) was started simultaneously with a controlled intravenous bolus injection of [11C]PE2I using an infusion pump (approximately 5 MBq/kg body weight). On the ECAT scanner, a 10 min transmission scan with rotating 68Ge rod sources was acquired in 2D mode prior to injection of the radioactivity. The emission scan was conducted in 3D mode. During the SIGNA scans, a series of MR sequences were obtained simultaneous with the PET data acquisition (10–65 min post start of scanning). Two sequences were of particular interest for this study: (1) a ZTE sequence (duration 153 s, 4 NEX, FOV 260 mm, slice thickness 1.4 mm, no slice gap, matrix 192 × 192, flip angle 0.8°) and (2) a 3D T1-weighted (TIw) brain sequence (gradient-echo, duration 272 s, 1 NEX, FOV 250 mm, slice thickness 1 mm, matrix 256 × 256, flip angle 12°, TI 450 ms).

### 2.4. PET image reconstruction

The PET data were reconstructed using ordered subsets expectation maximization (OSEM) using 6 iterations and 8 subsets for ECAT and 2 iterations and 28 subsets for SIGNA, applying a 4 mm Hanning and a 5 mm Gaussian post-filter, respectively. These reconstruction settings resulted in similar spatial resolution for ECAT and SIGNA data which was verified by measuring recovery coefficients, using a NEMA image quality phantom (Suppl. Fig. 1). Images were reconstructed into a 128 × 128 matrix with voxel dimensions of 2.06 × 2.06 × 2.43 mm (ECAT) or 2.34 × 2.34 × 2.79 mm (SIGNA). For the ECAT scans, AC was based on the 68Ge transmission scan (68Ge-AC). A prototype vendor-implementation of ZTE-MRAC was used for SIGNA AC (Duetto PET Reconstruction Toolbox MP26 [16–18]), which is the same version as we have evaluated before [19,20]. All other appropriate corrections for quantitative image reconstruction were applied as included in the scanner software.

### 2.5. Data analysis

PET images were corrected for motion using frame-by-frame motion correction, with each frame co-registered to the previous frame starting with an early summed image (0–3 min) as a reference. This 0–3 min image was also used for rigid co-registration of the 3D TIw MR scan to the PET scan. For this purpose, we used in-house developed software in Table 1

<table>
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<tr>
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<tr>
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<tr>
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</tr>
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<td>4.3</td>
</tr>
<tr>
<td>NEMA resolution axial at 1 cm (mm)</td>
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<td>5.3</td>
</tr>
<tr>
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</tr>
<tr>
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<td>5.9 cpm/kBq</td>
<td>23.3 cpm/kBq</td>
</tr>
<tr>
<td>Crystals per PMT</td>
<td>8x8 Crystals on 4</td>
<td>4x3 Crystals on 3x2</td>
</tr>
<tr>
<td></td>
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</tr>
<tr>
<td>TOF resolution (ps)</td>
<td>n.a.</td>
<td>390</td>
</tr>
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1 not applicable.
3. Results

3.1. Non-displaceable binding potential

For striatal BP_{ND} estimates, on average a moderate negative relative bias was found at the VOI level, which was significant for putamen and striatum as a whole (Fig. 1a). However, the SD of the relative bias was substantial, particularly for caudate. Correlations and orthogonal regression lines demonstrated a high agreement between BP_{ND} estimates based on ECAT and SIGNA scans (Fig. 1b). The correlations were close to one but somewhat lower for caudate (r ≥ 0.93).

Mean parametric BP_{ND} images for ECAT and SIGNA are displayed in Fig. 2a and b. A substantial negative absolute bias was found in subcortical regions in SIGNA images, particularly in the striatum (Fig. 2c). However, cortical areas near the soft tissue-bone interface displayed a slight positive absolute bias, above all in the posterior section of the brain (Fig. 2c). The T-maps indicated a significant negative absolute bias bilaterally in the temporal lobe (Fig. 2e).

3.2. Relative delivery

A significant positive relative bias was found for posterior cortical regions, whereas striatum and limbic regions showed a significant negative relative bias (Fig. 3a and Table 2). Anterior cortical areas and whole brain had an average relative bias close to zero, but ACR had the highest SD. The R1 estimates of SIGNA and ECAT showed generally a strong correlation (r > 0.90), although somewhat lower for posterior cortical regions (0.85), see Fig. 2b and Table 2. The slopes of orthogonal regression lines were close to identity, except for posterior cortical regions (non-significant).

Mean parametric R1 images for ECAT and SIGNA are shown in Fig. 4a and b. An evident negative absolute bias was found in the anterior section of the brain and a substantial positive bias in the posterior cortical regions (Fig. 4c). The T-maps revealed a positive bias bilaterally over the auditory cortex, which was most evident on the left side of the brain (Fig. 4d). A significant negative bias was noticed bilaterally in the anterior and temporal lobe (Fig. 4e).

3.3. Evaluation of time-activity curves

Mean SIGNA and ECAT based TACs expressed in SUV are presented for various brain clusters in Fig. 5a. Generally, there was a slightly larger inter-subject variation for SIGNA TACs compared to ECAT, except for striatum. Mean relative bias over time (Fig. 5b) was within ±10% for most brain clusters, but for posterior cortical regions, mean bias decreased from circa 20 to 10%. Striatum showed a nearly constant negative bias of about 5%. In contrast, the other brain clusters exhibited a bias pattern with a negative trend over time. When considering striatal regions, a relatively small and similar bias over time was found for putamen, while caudate demonstrated an increasing trend from circa −10 to 0% (Suppl. Fig. 2). Overall, the inter-subject variation of the TACs was consistent throughout the scan for all brain clusters with small fluctuations, mainly at the early time points.

4. Discussion

4.1. Significance

This work comprises the first quantitative brain study comparing PET/MR and stand-alone PET using state of the art MRAC in a clinical setting. Our main objective was to investigate whether quantitative outcome parameters of a dynamic [{\textsuperscript{11}}C]PE2I examination acquired on a hybrid SIGNA PET/MR scanner were equivalent to those from a dedicated ECAT PET examination, at both regional and voxel-level. For this purpose, we used a unique imaging dataset comprising a cohort of patients having both a scan on an integrated PET/MR a and a stand-alone...
PET system, using the latter as reference system. Today commercial stand-alone PET systems are not produced anymore and are mostly replaced by hybrid systems, combining PET with CT or MR. However, the results of this study are still important as $^{68}$Ge/$^{68}$Ga transmission sources for AC are generally regarded as the gold standard for quantitative brain PET imaging. It is also relevant to compare a PET/MR system against a stand-alone PET scanner to disentangle the summed effects of the differences in the scanner’s technology, geometry, and additional hardware in a dual system. These findings may be useful for further development of clinical PET/MR systems and applications, or alternatively, interpretation of data from a retrospective view.

### 4.2. Quantitative outcome measures $BP_{ND}$ and $R_1$

The VOI-based analysis showed a significant underestimation of striatal $BP_{ND}$ values for SIGNA. However, this average bias of 7–8% is close to the reported test–retest reproducibility of 4–9% for $[^{11}C]$PE2I $BP_{ND}$ [37]. For $R_1$, the VOI-based analysis of $R_1$ demonstrated significant underestimations of 6–9% on average in subcortical regions and overestimations of about 9% in posterior cortical regions when using SIGNA. In contrast, the mean differences between scanner systems were small for anterior regions and whole-brain (0.5–2%) and non-significant. Generally, a test–retest reproducibility of less than 10% is often used as a marker for clinical brain PET applications with various tracers. Thus, the biases found here, although significant in several cases, are within the range of a favorable reproducibility of $R_1$ estimates.

Voxel-wise analysis mainly confirmed the VOI-based analysis. Unexpectedly, voxel-wise analyses disclosed also bilaterally significant clusters in the temporal lobe, indicating underestimations of $BP_{ND}$ and $R_1$ for SIGNA. Further evaluations indicated that the underestimations in the temporal lobe might be caused by the used noise protecting headphones. Voxel-wise analyses of parametric $R_1$ also revealed few clusters with significant positive relative bias for SIGNA, particularly an overestimation in the parts of the auditory system, which was most likely activated by acoustic MR background noise. Both issues will be discussed in more detail in sections below.

To our knowledge, two other studies reported a comparison of a PET/MR system against a stand-alone PET scanner using $^{68}$Ge-AC. In the first case an integrated PET/MR (Siemens Biograph mMR), applying Dixon-MRAC and CT-AC, was compared with a stand-alone PET using $^{68}$Ge-AC (GE Advance) in a dynamic PET brain study with five healthy subjects [38]. As a PET tracer $[^{11}C]$Verapamil was used which required arterial blood sampling due to lack of a proper reference region. A serious systematic underestimation of the rate constants describing plasma-to-brain tissue ($K_{v}$) and clearance from tissue-to-plasma ($k_{2}$) as well as the distribution volume ($V_{T}$) was found when using Dixon-MRAC, for whole-brain grey matter ($–33$, $–19$ and $–18$%), insula ($–30$, $–27$ and $–5$%), and partial superior lobe ($–36$, $–23$ and $–16$%). When using CT-AC the bias was reduced with about 50%, but in several cases still over 15%. The authors concluded that the systematic differences between scanners precluded the combination of data from the PET/MR and stand-alone PET systems in mono- or multicentric studies without proper scaling adjustments.

The other case was an evaluation of a sequential PET/MR (Philips Ingenuity TF) with a vendor-provided MRAC based on image segmentation and a dedicated PET device (Siemens ECAT EXACT HR+) using $^{68}$Ge-AC in an $[^{18}F]$fluorodeoxyglucose (FDG) whole-body study with 13 patients [39]. A semi-quantitative evaluation was conducted using SUV as outcome parameter. Cerebellum was the only evaluated brain region and the observed relative mean bias was $12 \pm 4\%$. As TACs mostly reflect blood flow during the first 4 min, the SUV values within this time frame are approximately comparable to FDG tracer distribution [36]. In our work we found a relative bias of on average $7.62 \pm 0.50\%$ which is somewhat lower but with a larger inter-subject variation. Schramm et al. [39] concluded that their PET/MR system provided a reasonable accuracy but that there was still room for improvement. Due to the
interindividual segmentation quality, deviations of more than 20% could occur, which also was observed in our study for posterior cortical regions.

Comparisons with other studies should be done with care as differences in study populations, imaging systems, tracer kinetics, data acquisition, reconstruction and analyses are evident. However, these comparisons also indicate that the newest generation of MRAC methods cause that the differences between PET/MR systems and stand-alone PET has diminished.

4.3. Evaluation of time-activity curves

The causes of the differences in quantitative results between SIGNA and ECAT are not obvious and therefore the VOI TACs were further explored. Roughly, the TACs indicated a significant positive SIGNA bias in posterior cortical regions and a significant negative bias in limbic areas and striatum. The combination of time-varying bias in SUV values for the target VOIs and the reference region (cerebellum) explains in part the noticed bias in R1 and BPND estimates. The greater the difference between the TACs of the target and reference region, the greater the magnitude of the bias. Further, an overall positive SUV bias over time for the target region generally led to a positive bias in R1 and vice versa. The same pattern was found for striatal BPND estimates.

The bias between SIGNA and ECAT was time-varying (Fig. 5), which agreed with earlier findings. One reported hypothesis is that various MRAC methods are affected differently by the radioactivity distribution in the brain at different time points [38,40,41]. In our previous work, we also found that various MRAC methods compared to 68Ge-AC cause different patterns of bias over time and between VOIs on a SIGNA PET/MR scanner [20]. To disentangle the effects of AC from other sources of bias, we re-analysed in part the data of our previous publication (Suppl. Fig. 3) to enable a comparison with Fig. 5 in the current paper. In Suppl. Fig. 3, the SIGNA data, reconstructed with ZTE-MRAC, exhibited a consistent positive bias over time for all VOIs compared to SIGNA data reconstructed with 68Ge-AC. This pattern is distinctively different from what is presented in Fig. 5. Furthermore, variability between subjects was much smaller when only considering differences in AC methods (Suppl. Fig. 3) than when comparing the two different scanners (Fig. 5). Based on this ad-hoc analysis, we concluded that the estimated bias over time was not only caused by the differences in AC between SIGNA and ECAT.

A possible source of time-varying biases could be disparities in the implementation of scatter correction between both devices. The scanner’s geometry and existing attenuation material will affect scatter...
fractions. It is evident that SIGNA and ECAT differ in geometry, for example, because of different bore sizes, ring sizes, and the presence of coils in the SIGNA FOV. Although the scanner bed and MR coils are calibrated to the same dose calibrator. For ECAT, the calibration was more than 3% off. The SIGNA PET system is cross calibrated quarterly as recommended by the manufacturer. The linearity of both scanners was verified during acceptance testing. For SIGNA, this was annually in the temporal lobe for both BP
1
and BP
2
images [36] as both, to some extent, reflect cerebral blood flow. Equivalent to our results, Chonde et al. [46] demonstrated a significant bilateral cluster with increased FDG uptake (about 9%) when comparing between MR and PET imaging in a hybrid PET/MR system. Further, FDG images display non-specific glucose uptake, which corresponds well with [\(^{11}\)C] PE2I R
1
images [36] as both, to some extent, reflect cerebral blood flow. The voxel-wise analysis revealed a significant negative bias bilaterally in the temporal lobe for both BP
ND
and R
1
, which appeared to be related to the use of headphones. During the PET/MR investigation, the patients wore earplugs covered by headphones as protection from the MR background noise. The MRAC-map used for the reconstruction of the PET data does not consider this flexible hardware, therefore, headphones are potentially a source of error in quantitative PET data analysis. To confirm our findings, we investigated cases with and without headphones and their impact on SUVs in an ad-hoc experiment using NEMA 46-311426P1 and RSD striatal (RS-900T) phantoms on a Discovery MI PET/CT scanner (GE Healthcare, Milwaukee, WI, USA). A reduction in SUV of approximately 4% was found in the left and right temporal lobes when scanning with headphones. It should be noted that this experiment demonstrated additional attenuation due to the

4.5. Effects of flexible hardware

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ND
and R
1
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Table 2: R
1
: Mean accuracy (relative bias), precision (SD of bias) and Spearman correlation coefficient (r) for all brain clusters. Additionally, slope and intercept of orthogonal regression lines are given.

<table>
<thead>
<tr>
<th>VOI cluster*</th>
<th>Bias (%)</th>
<th>SD</th>
<th>r</th>
<th>Slope</th>
<th>Intercept</th>
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<tr>
<td>ACR</td>
<td>–0.48</td>
<td>3.86</td>
<td>0.91</td>
<td>0.84</td>
<td>0.12</td>
</tr>
<tr>
<td>PCR</td>
<td>9.30*</td>
<td>3.81</td>
<td>0.85</td>
<td>1.36</td>
<td>–0.17</td>
</tr>
<tr>
<td>STR</td>
<td>–8.87*</td>
<td>3.56</td>
<td>0.98</td>
<td>0.91</td>
<td>–0.01</td>
</tr>
<tr>
<td>LR</td>
<td>–6.11*</td>
<td>2.74</td>
<td>0.96</td>
<td>1.18</td>
<td>–0.20</td>
</tr>
<tr>
<td>WB</td>
<td>1.32</td>
<td>2.51</td>
<td>0.97</td>
<td>1.04</td>
<td>–0.02</td>
</tr>
</tbody>
</table>


Fig. 3. a) Relative bias of R
1
estimates for all observations within each VOI cluster. The dashed line in red indicates the mean; b) Correlation analysis between R
1
estimates based on SIGNA and ECAT scans for different brain clusters. Solid blue lines represent orthogonal regressions, and the dashed ones the line of identity. ACR: anterior cortical region, PCR: posterior cortical region, STR: striatum, LR: limbic region, WB: whole brain gray matter. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)
headphones but did not accurately assess the bias related to our results.

Previously, Büther et al. [47] investigated in a clinical setting the impact of MRI-safe headphones. They used FDG as tracer and compared sessions with and without headphones on a PET/MR (Siemens 3T Biograph mMR) and PET/CT (Siemens Biograph mCT), respectively. It was found that regional SUV values decreased by 8 to 11% in brain regions

Fig. 4. Mean R1 images for a) ECAT and b) SIGNA, c) absolute bias (SIGNA minus ECAT), and d,e) two-sample t-tests using SPM, showing voxel-wise differences between SIGNA and ECAT (p < 0.005). The red arrows in d) indicate the auditory cortex, on the most evident side. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Fig. 5. Time activity curves for all brain clusters: a) Mean SUV for ECAT (blue dots) and SIGNA (red dots) and b) mean relative bias of SIGNA (green dots) with standard deviation (shaded areas), ACR anterior cortical regions, PCR posterior cortical regions, STR striatal regions, LR limbic regions, CER cerebellum and WB whole brain grey matter. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)
close to the headphones, such as cerebellum, temporal lobe, and occipital cortex, but to a smaller degree in remote areas, such as basal ganglia (5%), frontal lobe (4%) and parietal lobe (3%). In another PET/MR study (Siemens 3T MR-Brain prototype) headphones were compared with in-earphones using a brain phantom filled with an 18F solution [48]. Like the previous study, an underestimation of 10% was found near the headphones due to improper AC. However, this negative bias was less than 1% when using in-earphones. Ferguson et al. [49] explored the effects of MR-safe headphones using a FDG phantom study and one human brain scan on a PET/MR scanner (Siemens 3T Biograph mMR). Activity concentrations measured in the entire phantom were reduced by 6.5% with headphones on compared to headphones off. In the human brain, the activity concentration was substantially reduced in regions near the headphones by as much as 14% in the inferior temporal and cerebellar cortex.

Based on these reports, we conclude that MR-compatible headphones are a significant source of attenuation and should be taken into account when conducting quantitative PET brain studies on a PET/MR system. One option is to visualize the headphones with MRI and correct the individual MRAC maps by simple segmentation-based techniques [50]. Both ZTE and UTE can visualize solid materials with short relaxation times. Another approach is to develop earphones such as described and analysed by Tellmann et al. [48] claiming a favourable sound absorption and ensured communication with the subject at a relative low cost. Further research on the use of headphones has been initiated but will be reported elsewhere.

4.6. Limitations

A limitation of the present work is that the results are based on only nine subjects from a heterogeneous group of patients with Parkinsonism. However, our sample size is in line with similar methodological studies. Two [11C]PE2I PET examinations of about 90 min in different environments were very demanding for this category of patients, and consequently, it might have influenced the scans’ repeatability. Furthermore, the ZTE-MRAC used in the present work is still undergoing continuous development [51]. For example, in this prototype, there was still a classification in the sinu regions. At a later stage, the manufacturer proposed a correction to this tissue misclassification [52] and released an improved ZTE-MRAC [18]. SIGNA PET scans were acquired mostly within three months after the ECAT scans, but the interval was up to 5 months for some patients. However, it is not likely that disease progression affected R1 and B100 bias significantly, given the patient’s conditions and relatively short time between the scanning occasions.

5. Conclusions

Quantitative outcome parameters of brain [11C]PE2I PET scans acquired on a SIGNA PET/MR system demonstrated a relative bias of the same order as the expected test–retest reproducibility of [11C]PE2I B100 and R1 with an error margin of about 6–10% compared to a stand-alone ECAT PET scanner. It is anticipated that the accuracy will be further improved when appropriate corrections are made for the attenuation of the used MR-compatible headphones, particularly in regions close to this flexible hardware. Functional PET/MR studies need to consider a potential source of error in auditory brain areas due to MR background noise. Altogether, this PET brain study resulted in a greater knowledge of potential quantitative differences at voxel level between a PET/MR and stand-alone PET system, which is highly relevant when introducing clinical applications on a PET/MR system.

6. Availability of data and materials

João M. Sousa (joao.sousa@ul.se) is the corresponding author for the data used in this manuscript.

7. Authors’ contributions

Concept, design, ethical application: ML, LA and HA; Recruitment of patients: DN, LA; Acquisition of data: LA; Data analysis: JMS, ME, ML; Data interpretation: JMS, ML, LA; Draft of the manuscript: JMS, LA, ML; Review of the manuscript: all authors; Approval of the manuscript: all authors.

8. Ethics approval and consent to participate

All procedures performed in studies involving human participants were in accordance with the ethical standards of the institutional and/or national research committee and with the 1964 Helsinki declaration and its later amendments or comparable ethical standards. Informed consent was obtained from all individual participants included in the study.

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Declaration of competing interests

HA and ML have received research support and speaker fees from GE Healthcare. ME is a former employee of GE Healthcare.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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Appendix A. Supplementary data

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References


