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Image-Based 2D Re-Projection for Attenuation Substitution In PET Neuroimaging

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Abstract

Purpose: In dual modality Positron Emission Tomography (PET)/Magnetic Resonance Imaging (MRI), attenuation correction (AC) methods are continually improving. Although a new AC can sometimes be generated from existing MR data, its application requires a new reconstruction. We evaluate an approximate 2D projection method that allows offline *image-based* reprocessing.

Procedure: 2-Deoxy-2-[¹⁸F]fluoro-D-glucose ([¹⁸F]FDG) brain scans were acquired (Siemens HR+) for 6 subjects. Attenuation data were obtained using the scanner's transmission source (SAC). Additional scanning was performed on a Siemens mMR including production of a Dixon-based MR AC (MRAC). The MRAC was imported to the HR+ and the PET data were reconstructed twice: once using native SAC (ground truth); once using the imported MRAC (imperfect AC). The re-projection method was implemented as follows. The MRAC PET was forward projected to approximately reproduce attenuation-corrected sinograms. The SAC and MRAC images were forward projected and converted to attenuation-correction factors (ACFs). The MRAC ACFs were removed from the MRAC PET sinograms by division; the SAC ACFs were applied by multiplication. The regenerated sinograms were reconstructed by filtered back projection to produce images (SUBAC PET) in which SAC has been substituted for MRAC. Ideally SUBAC PET should match SAC PET. Via coregistered T1 images, FreeSurfer (FS; MGH, Boston) was used to define a set of cortical gray matter regions of interest (ROIs). Regional activity concentrations were extracted for SAC PET, MRAC PET, and SUBAC PET.

Results: SUBAC PET showed substantially smaller root mean square error than MRAC PET with averaged values of 1.5% versus 8.1%.

Conclusions: Re-projection is viable image-based method for application of an alternate attenuation correction in neuroimaging.

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Conflict of Interest

The authors declare that they have no conflict of interest.

Keywords

Brain; Image enhancement/restoration; Inverse methods; Nuclear imaging; Validation

Introduction

Quantitatively accurate Positron Emission Tomography (PET) necessitates, among other things, correction for attenuation and scatter. To effect such a correction requires information about the attenuation properties of the subject being imaged. In PET, an attenuation factor (AF) can be defined for every sinogram element (i.e., for each line of response [LOR]). The AF for an LOR is the fraction of events that survive attenuation along that LOR. Thus, the inverse of this is an attenuation correction factor (ACF) which, in the absence of stochastic noise, is a multiplicative factor that can be used to correct the number of events recorded along each LOR for attenuation. Attenuation properties can also be encoded in a μ -image in which each voxel identifies an object's specific attenuation at that point (typical units: cm^{-1}) and which can be converted to a set of ACFs. Attenuation correction is applied prior to or as part of the image reconstruction process.

In some cases, it is necessary or desirable to retrospectively re-perform an attenuation correction (attenuation substitution). This situation can occur if the original attenuation correction is faulty in some way and a better version is available. The recent development of dual modality PET/Magnetic Resonance Imaging (MRI) provides a situation in which attenuation substitution may be needed. Algorithms for producing improved attenuation images from MR data are rapidly evolving [1] and are being developed by both manufacturers and independent research groups [2–11]. Thus, improved attenuation images may become available at some time after the completion of a PET/MR scan. Additionally, attenuation information for a particular subject may be available from other sources, e.g. a subject undergoing a PET/MR scan may have had a X-ray Computed Tomography scan that could be converted to a 511-keV μ -image.

When a new attenuation correction is to be used with existing PET data, ideally the new μ -image along with the original emission sinograms and scanner normalization data would be provided as input to the reconstruction software. Typically, this would be accomplished by loading the new μ -image and, if it is not already present, the archived sinograms and normalization on to the scanner and using the manufacturer's software for reconstruction. Unfortunately, loading a new attenuation correction onto a scanner in such a way that it actually can be used to perform a reconstruction can be a difficult process and may require assistance from the manufacturer. Further, if only the image data exist (i.e., not the sinograms and normalization data) this method cannot be used.

An alternative image-based approach is to approximately recreate the required projection data (sinograms) by forward projecting the image data. Given an attenuation-corrected PET image, the original μ -image, and the new, substitute μ -image, this method, which is described in detail below, can be used to produce a new attenuation-corrected PET image in which the new attenuation has been substituted for the original. In this approach, we essentially recreate offline sinogram data as needed from image data using the 2D Radon

(forward projection) and inverse Radon (filtered backprojection as implemented in Matlab) transforms. We observe that these operations are only truly inverse operations when operating on a consistent data set. Noise-free data from a physically realizable activity distribution, properly corrected for attenuation and scatter, produce sinograms that obey a set of consistency conditions [12]. However, sinograms with an improper attenuation correction generally do not. In the following, the application of the forward projection operation, by its nature, always results in sinograms that are consistent. Thus, the recreated (consistent) emission sinograms are only approximations to the original (inconsistent) emission sinograms with attenuation correction errors [13].

In this work, we implement and evaluate a 2D reprojection procedure for image-based attenuation substitution. We start with images in which an MR Dixon-based attenuation correction [14–15], without a bone segment, is used to perform the original scanner-based attenuation correction. We then use 2D forward projection to substitute a source-based (Ge-68/Ga-68) measured attenuation correction and compare the results to the case in which the source-based attenuation correction was used in a full reconstruction using the acquired emission data and manufacturer supplied software. Although it is a side effect of this work, our goal is not to determine the accuracy of the Dixon-based MRAC on PET reconstruction; Dixon-based MRAC has been characterized adequately by other authors [16–17]. Instead, our purpose is to determine the accuracy of the attenuation substitution method. For this, we require not only an accurate attenuation correction, here represented by the SAC, but an attenuation correction with inaccuracies (MRAC) as well. To perform our evaluation, our “inaccurate” attenuation correction must, however, be “reasonable”. We use the Dixon-based MRAC, which we define to be reasonable since it is in common use.

Materials and Methods

Attenuation Substitution Overview

Attenuation substitution via re-projection requires image data only; it does not require access to scanner sinogram, normalization, or calibration data. The method used for this evaluation is illustrated schematically in Fig. 1. As input the procedure requires: (Image 1) the original (i.e. from the PET scanner using its attenuation correction) attenuation-corrected PET image; (Image 2) the scanner-based μ -image; and (Image 3) the substitute μ -image. Input Image 1 is forward projected resulting in a PET emission sinogram that includes the original attenuation correction, that is, each sinogram element is multiplied by an attenuation correction factor (ACF) measured or calculated at the time of the original reconstruction. The original μ -image (Image 2), is also forward projected ($FP(\mu)$) and converted to ACFs using the antilog operation: $ACF = \exp(FP(\mu))$. The scanner-based ACFs are removed from the PET sinogram (forward projection of image 1 in Fig. 1) by division resulting in uncorrected PET emission sinograms. The uncorrected sinogram is multiplied by the new substitute ACFs, generated from the substitute μ -image (Image 3) using the same procedure employed to generate the ACFs from image 2, and reconstructed using filtered back projection.

An alternative but equivalent approach would be to subtract the original μ -image from the substitute μ -image (Image 3 – Image 2 in Fig. 1), forward project the result and take the

antilog. The quantities produced by this set of operations are the ratios of new to old ACFs. These are then used to multiply the image sinogram. Although the alternative approach is computationally simpler, we used the method shown in Fig. 1 because of our interest in examining various steps in the process.

This image-processing task is treated as a 2D problem: each image plane is treated separately. Further, in the forward projection procedure, no attempt is made to match the original scanner-based sinogram binning.

Subjects

As part of a separate research project, a set of subjects who were classified either as cognitively normal or with mild cognitive impairment (MCI) received PET scans with 2-deoxy-2-[^{18}F]fluoro-D-glucose (^{18}F]FDG) as the tracer on a Siemens HR+ scanner. Six of these subjects, ranging in age from 53 to 92 y, were rescanned on the mMR PET/MR scanner. Subject demographics are listed in Table 1. All studies performed as part of this and associated projects were conducted under protocols approved by the University of Pittsburgh Institutional Review Board. All subjects signed informed consent documents prior to entry into the studies.

Data Acquisition

For their initial scan, subjects were injected with 268 MBq (nominal) of ^{18}F]FDG and then rested in a quiet dimly lit room during the tracer uptake phase. Subjects were moved into the HR+ scanner and a 10 min transmission scan was performed using the scanner's integrated 511 keV Ge-68/Ga-68 transmission sources. PET emission scanning commenced 35 minutes after tracer injection. PET data were acquired into 6 5 min frames spanning a total of 30 min. Subjects were then transferred to the mMR scanner. The MR-based attenuation correction employed by the mMR is implemented using the Dixon method [14–15]. Thus, the required “in phase” and “opposed” T1 spin echo images were acquired. The PET emission data acquired on the mMR were not used in this project. Subjects also received an MPAGE T1 scan using a Siemens Trio MR.

Image Generation

Figure 2 illustrates the images produced for this project. Emission data from the HR+ and the attenuation data from both the HR+ and the mMR were used, however, for the purposes of the current analysis the mMR emission data were not used. We performed several studies comparing 3 different types of attenuation corrected PET scans. The first PET image (SAC PET, middle row of Fig. 2) was generated using the HR+ PET emission data reconstructed on the HR+ using the native HR+ SAC. This represents the “ground truth”. The second PET image (MRAC PET, top row) was reconstructed on the HR+ using the HR+ PET data but using the mMR MRAC imported onto the HR+. This image represents the result of reconstruction with an inaccurate attenuation correction. It is the starting, i.e. input, PET image for the attenuation substitution method. The third PET image (SUBAC PET, bottom row) is the output of attenuation substitution. Ideally, the SUBAC PET images should be identical to the SAC PET images.

PET Using Source-Based Attenuation Corrections (SAC-PET)

The SAC PET images were reconstructed directly from the transmission and emission data acquired on the HR+ using the manufacturer's software. As part of the process, the set of 3D sinograms was rebinned into a 2D set using Fourier rebinning (FORE) [18]. Two different reconstruction methods were tested in this project, Direct Fourier Transform (DIFT, similar to filtered back projection) [19] and Ordered Subsets Expectation Maximization (OSEM) [20] using 4 iterations of 16 subsets. In each case, images were reconstructed into a $128 \times 128 \times 63$ grid with voxels sizes of $2.6 \times 2.6 \times 2.4$ mm (zoom factor of 2).

Reconstructions included corrections for dead time, decay, scatter, as well as attenuation. No apodizing or post-reconstruction filtering was performed for either of the reconstruction methods. For both methods, each data time-frame was processed separately and a single frame was produced by averaging the results over frames. For each reconstruction method, the SAC PET is used as a gold standard for comparison.

At the time of reconstruction, the scanner provided a μ -image with the same pixel size ($2.6 \times 2.6 \times 2.4$ mm, zoom factor 2) as the reconstructed PET image for use in the attenuation substitution procedure. The scanner also provided a zoom factor 1 μ -image with voxel size $5.7 \times 5.7 \times 2.4$ mm for use in the MRAC registration process described below.

PET Using MR-Based Attenuation Correction (MRAC PET)

The manufacturer-supplied mMR software generates an MR/Dixon-based attenuation correction (MRAC) by using the acquired MR in phase and opposed phase images to derive fat and water images and then to classify voxels into one of 3 segments: air, fat, or soft tissue. Each of these segments is assigned a fixed 511-keV attenuation value.

The MRAC PET images were reconstructed identically to the SAC PET except that the MRAC μ -image, acquired and produced by the mMR, was used to generate the attenuation and scatter correction. Reconstruction of the MRAC PET image required that the MRAC μ -image be exported from the mMR, aligned with the HR+ PET data and imported to the HR+. To be successfully used for attenuation correction, the HR+ requires that imported μ -images have a zoom factor of 1, which corresponds to a voxel size of $5.7 \times 5.7 \times 2.4$ mm. For the MRAC μ -image to be appropriate for attenuation correction of the HR+ PET data, the μ -image should include the HR+ head holder, which is in the field-of-view during the data acquisition. These requirements were met using the following procedure. The native, zoom-factor 1, source-based μ -image that was produced at the time of initial HR+ data reconstruction provided a reference image for alignment of the MRAC. The MRAC μ -image was exported from the mMR and then aligned to the reference image using the image fusion tool of PMOD (PMOD Technologies LLC, Zurich). The procedure was carried out by an initial manual alignment followed by an automated fine alignment via the normalized mutual information option within the PMOD fusion tool. The registered MRAC μ -image was smoothed in 3 dimensions with a 5 mm Gaussian filter to approximately match the HR+ resolution. PMOD was used to edit the native source-based μ -image to produce an image of just the head holder which was then inserted into the realigned and smoothed mMR-based MRAC μ -image. The net result of the process, illustrated in Supplemental Fig. 1, is an aligned MRAC μ -image that includes the HR+ head holder and which is resliced to the

required voxel size. The final image registration between the μ -images was deemed acceptable when no visually apparent misalignment could be observed. In practice, a careful initial manual alignment always resulted in an acceptable final alignment.

This image was used as the attenuation correction in a reconstruction of the PET data using the HR+ software, thus providing the MRAC PET image. In the analysis, the MRAC PET represents a PET image with an imperfect attenuation correction. It will be treated as if it were an original image (image #1 in Fig. 1) from, e.g., a dual modality PET/MR scanner. As in the case of the SAC PET image preparation, the scanner also produced an MRAC μ -image, with the same voxel size as the PET image, for use in the attenuation substitution process.

PET Using Substitute Attenuation Correction (SUBAC PET)

The attenuation substitution algorithm takes as input the MRAC PET, the μ -image encoding the MRAC (to be removed from the PET image), and the μ -image encoding the SAC (to be substituted). For this work, the μ -images with zoom-factor 2 were used. The attenuation substitution procedure was implemented in Matlab (The MathWorks, Inc.). The input images to the algorithm were converted from ECAT (Siemens HR+ format) to Matlab structures via PMOD. The procedure outlined here was followed independent of the method of original reconstruction (DIFT or OSEM).

The process was carried out as a series of 2D operations in that each image transaxial (Z) plane was treated separately. Each was forward projected over the 180 deg domain appropriate for PET using the “radon” function of Matlab with angular binning of 0.5 deg. Attenuation factors were generated from the forward projected μ -images by using the antilog operation. The MRAC ACFs were removed from the forward projection of the PET image by division, and the SAC-based attenuation factors were substituted by multiplication. The resulting projections with substitute attenuation correction were converted back to an image by filtered backprojection using the “iradon” (inverse Radon) Matlab function.

Evaluation

For each subject and reconstruction method, the image processing described above yields 3 PET images: SAC PET, the gold standard; MRAC PET, PET processed with a faulty attenuation correction, and SUBAC PET, the result of running the substitution process starting with MRAC PET as the input. Ideally SUBAC PET should match SAC PET. The evaluations described below are intended to both show the deviation of the SUBAC PET from the SAC PET and to illustrate the level of image improvement obtained with SUBAC PET compared to MRAC PET.

Ratio Images

To illustrate the performance of the method as a function of anatomical position, for each subject, ratio images were computed: MRAC PET / SAC PET and SUBAC PET / SAC PET. Each component image was filtered in 3 dimensions using a Gaussian kernel of 8 mm prior to division. Voxels with less than 25% intensity of the maximum-uptake voxel in the SAC

image were masked out of the final ratio image. Imperfections in the MRAC PET or SUBAC PET manifest in the form of deviations from a uniform value of 1 in the ratio images.

Regional Evaluation

For each subject, the separately acquired T1 MPRAGE image was registered to the PET images. FreeSurfer (<http://surfer.nmr.mgh.harvard.edu/>) was used to parcellate the image into FreeSurfer's default set of ROIs. The ROIs were then applied to the PET using PMOD to yield statistics for each ROI. Values for the mean activity in each FreeSurfer cortical gray matter region were tabulated.

For each subject, correlation plots were constructed showing SAC PET ROI activity versus MRAC PET activity and SAC PET activity versus SUBAC PET activity. For each, the Coefficient of Determination R^2 was calculated. Fractional error, F_{MRAC} , was calculated for each MRAC PET ROI as $F_{\text{MRAC}} = \frac{\lambda_{\text{MRAC}} - \lambda_{\text{SAC}}}{\lambda_{\text{SAC}}}$, where λ_{MRAC} and λ_{SAC} are the ROI

activities in MRAC PET and SAC PET respectively. A SUBAC PET fractional error F_{SUBAC} was calculated similarly but with λ_{SUBAC} , the ROI activity in the SUBAC PET ROI, in place of λ_{MRAC} in the above equation. Per subject relative (to SAC PET) bias and root mean square error (RMSE) over all ROIs was calculated. Averages of the quantities calculated for each subject were summarized as the average and standard deviation (population) across subjects.

Results

Figure 3 is an example ratio image from the DIFT reconstruction of subject b. Rows 1 and 2 show an orthogonal view for the MRAC PET divided by the SAC PET and SUBAC PET divided by SAC PET respectively. Images are shown using a blue (less than 1) – red (greater than 1) color scale in which values near 1 are white. The two ratio images are shown on the same color scale. For anatomical reference, the third row shows the subject's T1 MPRAGE MR image, which is in registration with the ratio images. It is apparent that the SUBAC PET yields a more accurate result than MRAC PET. We also observe as have others [16–17] that there is a spatial gradient associated with the MRAC PET. The gradient is the result of the lack of a bone segment in the MRAC and is not present in the SUBAC PET. All other ratio images for both the DIFT and OSEM reconstructions showed features similar to those apparent here.

Figure 4 shows the SAC PET versus MRAC PET (blue) and SAC PET versus SUBAC PET (red) for the FreeSurfer cortical ROI activity averages for all 6 subjects for the DIFT reconstruction. Each plot also displays the line of identity and the best least squares linear-fit lines to the two data sets. It can be seen that the SUBAC PET results show a stronger correlation than the MRAC PET results and lie closer to the line of identity.

These results are further illustrated by Fig. 5, which shows fractional error (F) histograms for the DIFT reconstructions for the of FreeSurfer cortical ROI activity. Importantly the width of the SUBAC PET histograms, which reflect RMSE are substantially reduced compared to MRAC PET.

Results are quantitatively summarized for the DIFT reconstruction with FreeSurfer cortical ROI activity in Table 2. MRAC PET and SUBAC PET results are compared for each subject. The first set of results columns in Table 2 compares the coefficient of determination (R^2) of the linear fits illustrated in Fig. 4. For MRAC PET R^2 ranges from 0.728 to 0.882 whereas for the SUBAC PET all values of R^2 are greater and range from 0.985 to 0.998.

Biases in Table 2 were calculated on a per subject basis as the mean value over all cortical ROIs of F_{MRAC} or F_{SUBAC} . On average, across subjects, biases for both MRAC PET and SUBAC PET were small, with an absolute value of less than 1%. However, the bias range was larger in the case of MRAC PET (-0.391 to 0.0583) than for SUBAC PET (-0.0164 to 0.00285). This point is related to the RMSE findings, which are much smaller for SUBAC PET (average 0.015) than for MRAC PET (average 0.081).

The results for OSEM reconstruction are similar to those for DIFT in that attenuation substitution method produces substantially better, i.e., more like the SAC PET, than MRAC PET. Table 3, which is laid out as is Table 2, summarizes the OSEM results.

Discussion

The procedure presented here makes use of the Radon (forward projection) and inverse Radon transform (filtered back projection) as implemented in Matlab. The new ACFs are substituted for the original ACFs between the application of these two inverse operations. After removal of the original attenuation factors and with the availability of the substitute ACFs, in principle, any reconstruction method could be used to produce the final PET image with substitute attenuation. However, as our purpose here is strictly to substitute one attenuation correction for another, we choose to use inverse operations to transform between image space and projection space and back. This has the desirable feature that, in the limit in which the substitute attenuation image becomes equal to the original attenuation image, the final PET image will match the original.

If the effects of scatter are neglected, the procedure is formally most correct for data that were initially reconstructed in 2D, without 3D rebinning, using the inverse Radon transform (i.e. FBP). In this case, the Radon transform applied to the PET image as the first step of the attenuation-substitution procedure approximately (see Introduction) undoes the initial reconstruction. However, in modern practice, in which 3D projection data is generally acquired, such simple methods are rarely used. In this work we evaluated the attenuation substitution procedure using data that were acquired in a 3D mode and rebinned into 2D projections using FORE, and then reconstructed using DIFT [19], an analytic method similar to FBP. The attenuation substitution procedure evaluated here does not explicitly address the fact that original image was reconstructed from rebinned data; instead a simply implemented 2D method was employed. Nevertheless, the final SUBAC PET images were accurate and a substantial improvement over the initial MRAC PET images. Taking us one step further from formal exactness, the method was evaluated starting with an initial OSEM reconstruction [20], which approximates a maximum likelihood expectation maximization [21] image. Once again good results were obtained, comparable to the results using the analytic reconstruction. In all cases the original on-scanner image reconstruction included

scatter correction [22] which depends on the attenuation distribution and, ideally, should be recalculated using the new attenuation correction. This effect was neglected in this work.

More sophisticated methods of reconstruction are available on modern scanners that we were not able to evaluate because of limitations of our HR+ scanner. Specifically, we did not test the attenuation substitution method with point-spread function (PSF) reconstruction, which is now offered by all major PET manufacturers. This method compensates for the blurring introduced by the imaging process and makes use of a spatially varying point spread function that is precisely measured and incorporated into the reconstruction engine by the manufacturer. Based on our OSEM results, we believe the attenuation substitution method can be used with benefit on images initially produced using PSF reconstruction. However, the extra processing involved in the attenuation substitution method, i.e., application of the Radon and inverse Radon transform introduces extra interpolations. We point out that in Fig. 2 there is a hint of reduced image sharpness in the SUBAC PET compared to the SAC PET that might be due to these effects and which would likely be more noticeable on a higher resolution starting image such as from PSF reconstruction. An approach to ameliorating this potential issue with high-resolution images would be to reduce the bin size used in the Radon process.

Thielemans et al. [23] have proposed an alternative approach to image-based attenuation substitution in which the effect of mismatched attenuation on the resultant PET image is characterized in terms of a series expansion in the μ error. They then specify an iterative method for correcting the final PET that provides reasonably accurate results in several test cases. To insure convergence, the authors note that the μ error cannot be too large. In contrast, the current procedure is not iterative and there is no formal issue of series convergence. However, in the current procedure, as the error in the attenuation increases, the regenerated sinogram (from the initial forward projection) is likely to increasingly diverge from the original sinogram, depending on the precise nature of the object geometry and μ error [13]. An approximation common to both methods is that scatter is not recalculated, but ideally should be.

Conclusions

Regeneration of 2D sinogram data through backprojection of previously reconstructed images provides an image-based capability for reprocessing of PET data with an alternative attenuation correction. The method provided substantially improved image accuracy in test cases where it was used to substitute an attenuation correction based on a “correct” source-based μ -image for an MR/Dixon-based μ -image with no bone segment.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Fig. 1.

Re-projection method, described in the text, requires image data only. Input images are shown at the top of the diagram and the resulting output image at the bottom. Images are forward projected to sinogram space. In the case of the μ -images, ACFs are calculated as $ACF = \exp(FP(\mu))$ where $FP(\mu)$ is the forward projection operation performed on the μ -image. The original ACFs are removed by division and the substitute ACFs applied by multiplication. The resultant sinogram is then reconstructed using filtered back projection.

Fig. 2. Sample images from subject b. The first column shows the Dixon MRAC μ -image (top) and the measured SAC μ -image (bottom). The remaining columns show orthogonal views (axial, sagittal, coronal) of the PET image generated using the attenuation corrections shown in the first column. **a** The top row of PET images were reconstructed on the HR+ scanner with DIFT using the Dixon MRAC and represent, for example, what might be acquired on the PET/MR scanner. **b** The second row of PET images were also reconstructed with DIFT on the PET scanner but using the native SAC. The bottom row of PET images was generated starting with the MRAC PET (top row) and applying the substitution method to replace the MRAC with the SAC. Ideally, the SUBAC PET should match the SAC PET. PET images and the SAC μ -image are oriented as acquired on the HR+ scanner. The MRAC has been realigned to match this orientation as described in the text.

Fig. 3.

Orthogonal view of ratio images for subject b: **a** MRAC PET/SAC PET; **b** SUBAC PET/SAC PET; and, **c** corresponding T1 MR. Ratio image color scale as displayed runs from 0.6 (blue) to 1.4 (red). Some regions external to the brain survived the 25% threshold cut, described in the text, and are visible in the ratio images. The images shown have been reoriented to AC-PC alignment.

Fig. 4.

Scatter plots for each of the 6 subject scans, initially reconstructed on scanner with DIFT, showing correlation between SAC PET and MRAC PET ROI averages (blue) and between SAC and SUBAC (Red) PET ROI averages. There is a blue and red point for each of the FreeSurfer cortical gray matter ROI. The dashed line is the line of identity. The blue and red dotted lines show the best linear fit to the SAC PET vs. MRAC PET and SAC PET vs. SUBAC PET respectively.

Fig. 5.

F-histograms (described in text) showing, for each subject scan, initially reconstructed with DIFT, a histogram over FreeSurfer cortical gray matter ROIs of PET fractional error using MRAC (blue) and PET fractional error using SUBAC (red). Fractional errors are computed with respect to the PET with SAC.

Table 1.

Subject Information

Subject ID	Sex	Age (Y)	[¹⁸ F]FDG Dose (MBq)	Cognition Status
a	M	90	275	Normal
b	M	87	258	Normal
c	F	53	268	MCI
d	M	92	270	Normal
e	M	58	275	MCI
f	M	90	264	Normal

MCI=Mild C

Table 2.

MRAC PET and SUBAC PET results for the cortical gray matter regions for the case of an initial DIFT reconstruction

ID	R^2		Bias		RMSE	
	MRAC	SUBAC	MRAC	SUBAC	MRAC	SUBAC
a	0.763	0.998	-0.0391	-0.00664	0.0987	0.0128
b	0.851	0.998	0.00947	-0.00643	0.0719	0.0109
c	0.728	0.998	0.0583	0.00257	0.114	0.00819
d	0.747	0.994	0.0032	-0.00285	0.0758	0.0132
e	0.832	0.996	-0.0118	-0.0124	0.0625	0.0167
f	0.882	0.985	0.0196	-0.0164	0.0623	0.0269
Avg	0.80	0.99	0.0066	-0.0070	0.081	0.015
SD	0.063	0.0051	0.033	0.0067	0.021	0.0066

Table 3.

MRAC PET and SUBAC PET results for the cortical gray matter regions for the case of an initial OSEM reconstruction

ID	R^2		Bias		RMSE	
	MRAC	SUBAC	MRAC	SUBAC	MRAC	SUBAC
a	0.778	0.995	-0.0369	-0.00172	0.0963	0.0151
b	0.866	0.993	0.00588	-0.00743	0.0661	0.0236
c	0.722	0.996	0.0581	0.00361	0.114	0.0123
d	0.785	0.991	0.000814	-0.00393	0.0719	0.0190
e	0.83	0.996	-0.0076	-0.00722	0.0655	0.0135
f	0.877	0.982	0.019	-0.016	0.0628	0.0280
Avg	0.81	0.99	0.0065	-0.0054	0.079	0.019
SD	0.059	0.0053	0.031	0.0066	0.021	0.0062