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FULL PAPER

Partial-volume effect correction in positron emission tomography brain scan image using super-resolution image reconstruction

¹T MEECHAI, ²S TEPMONGKOL and ³C PLUEMPITIWIRIYAEWJ, PhD

¹Department of Biomedical Engineering, Chulalongkorn University, Bangkok, Thailand

²Division of Nuclear Medicine, Department of Radiology, Faculty of Medicine, Chulalongkorn University, Bangkok, Thailand

³Department of Electrical Engineering, Chulalongkorn University, Bangkok, Thailand

Address correspondence to: Assistant Professor Charnchai Pleumpitiwiriyaewj

E-mail: cpluem@yahoo.com

Objective: The partial-volume effect (PVE) is a consequence of limited (*i.e.* finite) spatial resolution. PVE can lead to quantitative underestimation of activity concentrations in reconstructed images, which may result in misinterpretation of positron emission tomography (PET) scan images, especially in the brain. The PVE becomes significant when the dimensions of a source region are less than two to three times the full width at half maximum spatial resolution of the imaging system. In the present study, the ability of super-resolution (SR) image reconstruction to compensate for PVE in PET was characterized.

Methods: The ability of SR image reconstruction technique to recover activity concentrations in small structures was evaluated by comparing images before and

after image reconstruction in the NEMA/IEC phantom (Washington, DC), in the Hoffman brain phantom and in four human brain subjects (three normal subjects and one atrophic brain subject) in terms of apparent recovery coefficient (ARC) and percentage yield.

Results: Both the ARC and percentage yield are improved after SR implementation in NEMA/IEC phantom and Hoffman brain phantom. When tested in normal subjects, SR implementation can improve the intensity and justify SR efficiency to correct PVE.

Conclusion: SR algorithm can be used to effectively correct PVE in PET images.

Advances in knowledge: The current research focused on brain PET scanning exclusively; future work will extend to whole-body imaging.

Positron emission tomography (PET) is a nuclear medicine imaging technique used to image the physiological function of organs. It has quite a limited spatial resolution compared with CT scan or MRI. The resolution of PET imaging is affected by the inherent physical effects such as detector size, positron range and non-collinearity.¹ Other causes that can degrade PET resolution include the reconstruction algorithm and patient motion. A consequence of the limited spatial resolution of PET scans is the partial-volume effect (PVE), which is a phenomenon where reconstructed activity concentrations are lower than actual activity concentrations. This occurs in lesions where dimensions are less than three times the full width at half maximum (FWHM) of the system spatial resolution.^{1,2} In order to accurately recover image-derived activity concentrations in source regions such as small tumours and small structures in the brain, PVE correction is required. Without such corrections, underestimation of activity concentrations may lead to mischaracterization of tissue structures and,

potentially, misdiagnosis. There are several methods for PVE correction, each of which provides different effective results. Some PVE correction techniques use MRI image for PVE correction.³ In these cases, MRI provides useful anatomical data to define the region boundary. Thus, the efficiency of PVE correction depends on PET/MRI image registration and MRI segmentation.⁴ The appropriate segmentation technique should be considered if MRI image is used in the PVE correction technique.⁴

The recovery coefficient (RC) method⁵ assumes that tumour volume and shape of the part affected by PVE is known. The RC is a very simple method for PVE correction. This method uses a pre-calculated correction factor, which is the relationship between the object size (diameter) and the ratio of restored activity and the true activity. RC factors depend on spatial resolution of the imaging system, tumour volume, tumour shape and image sampling. This method can be applied to lesion sizes of down to

1.5 × FWHM of a PET scanner. RC can improve only the activity value but not the image quality.

The geometric transfer matrix method^{3,6–9} has been successfully used in PET brain imaging. The method assumes that the structure has uniform uptake and the structure's volume is known. This technique calculates the transfer coefficient to each compartment from anatomical data. Each compartment of interest will be blurred by the point spread function (PSF) of the imaging system. The transfer coefficient forms a matrix representing the fraction of true activity from each brain region observed in any given set of regions of interest (ROIs). This matrix can be inverted to correct for PVEs, independent of the tracer concentrations in each tissue component.

The deconvolution method¹⁰ also requires prior knowledge of the tumour volume and shape. This technique uses a PSF to recover the spatial resolution of the PET image. This technique requires prior knowledge about the structure volume. The PVE correction by deconvolution has been suggested to estimate the spillover effect caused by PSF. The iterative deconvolution technique is used to estimate the PVE-corrected image. At each iteration, the image that would be obtained is calculated and compared with the image that was acquired by the deconvolution process. Once the comparison is complete, a new estimate image can be obtained when the difference between the calculated image and the obtained image is minimized. This technique may amplify the level of noise and should not be applied to PET images that contain lesion sizes of <1.5 times of FWHM.

The partition-based correction¹¹ method assumes that the uptake of all compartments surrounding the tumour is known and the tumour has uniform uptake. MRI data are needed to increase the resolution of the PET image by this method.^{11–14} This method assumes that the true activity distribution can be segmented into each compartment with a known uniform uptake, except for the interested compartment. The PVE-corrected image by this method can be obtained by first convolving the image of each compartment, scale to its uptake value by the PSF and then summing the resulting image. The activity distribution of the interested region is obtained by the subtraction of the actual image from the PVE-corrected image.

The multiresolution approach¹⁵ needs anatomical data from CT or MRI to reconstruct the PVE correction image. Details from high-resolution image (CT or MRI) are extracted and integrated in a low-resolution PET image. Discrete wavelet transforms of a high-resolution image (CT or MRI) and PET are performed to identify the level of spatial resolution. Details at the level of resolution present in the CT or MRI image are incorporated into the PET image on the basis of the assumed correlation between gray levels in the two types of images. This technique can correct both the radioactivity concentration and the quality of the image.

The fitting method¹⁶ assumes the tumour has a spherical shape and uniform uptake. The observed image is the result of the convolution of this sphere and the PSF. The unknown

parameters of the model will be estimated by minimizing an objective function characterizing the quality of the fit and penalizing unrealistic solutions. This technique depends on the local PSF of the interested area. The unknown parameters can be acquired from object fitting model and then PVE correction image can be solved.

The modeling PVE during reconstruction or anatomic maximum *a posteriori* method¹⁷ uses a “maximum *a posteriori*” approach to incorporate anatomical data into the reconstruction process. Anatomical data can preserve the edges between different compartments. The MRI image is used to derive a model of tissue compartment and compensate for the tissue fraction effect. The PVE induced from tissue fraction effect can be solved. The corrected PET image shows strong contour edges from MRI.

Many of the current techniques are limited owing to the fact that knowledge of tumour volume or anatomical data is necessary, and among the techniques, some require the lesion to be spherical. To obviate the need for anatomical data for PVE correction, we selected an image reconstruction technique known as iterative-interpolation super-resolution image reconstruction proposed by Irani and Peleg.¹⁸ This technique is known to improve the spatial resolution and contrast ratios of PET images;¹⁹ however, the ability of this technique to correct PVE has yet to be documented. Super resolution (SR) is an image-processing technique that can reconstruct a high-resolution image using a set of low-resolution images with different points of view (POVs) by shifting and/or rotating the object. With this technique, there are some disadvantages that are impractical for use in humans. Firstly, scan duration is increased, and secondly, the geometric transforming method that is performed by moving the object is not suitable in patient studies. The later modified SR technique by Chang et al²⁰ that acquires different POVs by shifting the centre of the reconstruction grid without the need to move the patient is used here.

METHODS AND MATERIALS

Super-resolution method

The mathematic framework of the Irani and Peleg¹⁸ algorithm for solving SR image by iterative reconstruction is as follows.

The process of SR starts with the estimation of the high-resolution image (SR-implemented image) from the set of low-resolution images (original PET images). Then, it is updated based on the average of the downsampled acquisitions, which are shifted to a common reference frame:

$$f^{(0)} = \frac{1}{K} \sum_{k=1}^K T_k^{-1} \{g_k \uparrow S\} \quad (1)$$

where $f^{(0)}$ is the initial high-resolution image guess; g_k is the k th observed low-resolution image frame; $\uparrow S$ is the upsampling operator from low-resolution to high-resolution image, by increasing the matrix size of the original PET image; T_k^{-1} is the inverse geometric transform of the k th acquisition; K is the number of PET images used in SR image reconstruction.

Figure 1. Positron emission tomography images of the NEMA/IEC phantom (Washington, DC). The original image (left) and super-resolution-implemented image (right). (a) The whole image of NEMA/IEC phantom and (b) the zoom image of 13-mm sphere size.

Next, the low-resolution image after n iteration $g_k^{(n)}$ can be estimated from the high-resolution image $f^{(n)}$ with the equation:

$$g_k^{(n)} = \left(T_k \{ f^{(n)} \} \times h \right) \downarrow S \quad (2)$$

The process includes shifting the image f to the k th POV, blurring the image by convoluting with the PSF, h , and then down-sampling the image, $\downarrow S$, by taking image averaging and adding noise. T_k is the geometric transform of the k th acquisition. The point spread (or function), h , was measured. The test is based on the NEMA NU2-2007 spatial resolution test. We used three point sources of fluorine-18 (^{18}F), each with dimensions of <1 mm in both transaxial and axial directions. Sources are prepared using foam beads 1–2 mm in diameter soaked in an ^{18}F -containing solution. The sources are placed in three positions:

- (1) at $x = 1$ cm and $y = 0$ cm [1 cm vertically from the centre of rotation to represent the centre of the field of view (FOV),

but positioned to avoid any possible inconsistencies in results at the very centre of the FOV]

- (2) at $x = 0$ cm and $y = 10$ cm

- (3) at $x = 10$ cm and $y = 0$ cm.

The activity of the sources should be such that when all the sources are in the FOV, the percentage of dead time lost is random and $<5\%$ of the total event rate. Typically, this can be obtained with an activity of the order of 1 MBq. The activity concentration of the starting radioactive solution should thus be approximately 1000 MBq ml^{-1} (or 27 mCi ml^{-1}) or less.

At least 100,000 counts were acquired for each response function to account for reduced count statistics after correction for random and scatter events. The reconstruction method was the ordered-subset expectation maximization (OSEM) with a ramp filter; no further smoothing or any image processing was applied. The PSF was analysed. The blurring function to be used in the SR algorithm was calculated from the PSF thus derived.

To obtain a better estimate of the image, f , the difference between the low-resolution data g_k and $g_k^{(n)}$ was used to update the higher resolution image, f^n . The next iteration f^{n+1} of the high-resolution image follows the equation:

$$f^{n+1} = f^{(n)} + \frac{1}{K} \sum_{k=1}^K \left(T_k^{-1} \left\{ \left(g_k - g_k^{(n)} \right) \uparrow S \right\} \right) * p \quad (3)$$

The difference between g_k and $g_k^{(n)}$ are upsampled using the upsampling operator, $\uparrow S$, to achieve the high-resolution image and then moved to common reference frame, T_k^{-1} and average overall k frame acquisition. The symbol $*p$ is a sharpening kernel that is the inverse of the blurring kernel. The iterative process is continued until the difference between g_k and $g_k^{(n)}$ is minimized. The minimizing function $e^{(n)}$, which is the difference between g_k and $g_k^{(n)}$, is calculated from Equation (4). The SR process will stop when $e^{(n)}$ is minimized:

$$e^{(n)} = \sqrt{\sum_k \sum_{(x,y)} \{ g_k(x,y) - g_k^{(n)}(x,y) \}^2} \quad (4)$$

where $g_k^{(n)}$ is the lower resolution image obtained by applying the simulated imaging process of $f^{(n)}$; $f^{(n)}$ is the approximation

Table 1. Activity concentration and apparent recovery coefficient (ARC) in different sphere sizes

Sphere size (mm)	True activity concentration (Bq ml^{-1})	Activity concentration in original image (Bq ml^{-1})	Activity concentration in SR image (Bq ml^{-1})	ARC (original image)	ARC (SR image)
10	51,800	21,460	38,480	0.41	0.74
13	51,800	29,600	39,960	0.57	0.77
17	51,800	41,810	45,410	0.81	0.87
22	51,800	45,510	45,510	0.88	0.88
28	51,800	51,060	52,910	0.99	1.02
37	51,800	51,430	52,170	0.99	1.01

SR, super resolution.

Figure 2. The percentage yield in different sphere sizes of NEMA/IEC phantom study (Washington, DC).

of f obtained after n iterations; and g_k is the k th observed low-resolution image.

NEMA/IEC phantom study

In our study, we used the NEMA/IEC phantom (Washington, DC) to investigate the efficiency of SR algorithm to improve the activity in each sphere. In the phantom, there are six different sphere sizes with diameters of 10, 13, 17, 22, 28 and 37 mm, respectively, with a sphere-to-background activity concentration ratio of 10:1. The activity concentration of $1.4 \mu\text{Ci cm}^{-3}$ was filled in every sphere. The background of the phantom was filled with the activity concentration of $0.14 \mu\text{Ci cm}^{-3}$.

The phantom was positioned at the centre of the Siemens Biograph 16 PET/CT scanner (Siemens Healthcare, Erlangen, Germany). Data were acquired in three-dimensional (3D) mode for 10 min per one data acquisition. In these data acquisition phases, an original PET image (low-resolution image) with a matrix size of 256×256 pixels was reconstructed in the

51.2-cm FOV. Consequently, the reconstruction was performed using the OSEM algorithm with four iterations and eight subsets. Attenuation correction was performed by CT mapping. Scatter and random corrections were compensated automatically via Siemens software. These corrections were also applied to Hoffman brain phantom and human studies. As for the SR image, the eight different low-resolution images with 256×256 -pixel matrix size were generated by shifting the centre of the reconstruction grid along the x or/and y axes within subpixel distance to generate the images in different POVs following the technique of Chang *et al.*²⁰ Then, the same reconstruction algorithm as described above was used for the SR protocol.

Hoffman brain phantom study

The Hoffman brain phantom is used to simulate activity distribution in the brain PET study before applying the algorithm to the patient. An uptake ratio of 4:1 is observed for normal grey and white matter. A $1.4 \mu\text{Ci cm}^{-3}$ ^{18}F -fluorodeoxyglucose (^{18}F -FDG) activity concentration was used. The phantom was positioned at

Figure 3. Hoffman brain phantom images and line profiles across the right and left caudate regions of (a) the original image and (b) the super-resolution (SR) image.

Figure 4. Improvement of regional counts (partial-volume effect correction) after reconstruction with the super-resolution (SR) algorithm: (a) original image and (b) SR image (arrows).

the centre of the PET scanner's FOV. Data were acquired in 3D mode for 15 min. In these data acquisition phases, a low-resolution image with a matrix size of 336×336 pixels was reconstructed. Consequently, the reconstruction algorithm was performed using the OSEM algorithm with 4 iterations and 24 subsets. For SR implementation, the original PET image was acquired in 3D mode for 15 min. After data acquisition, six different low-resolution images with a 336×336 -pixel matrix size were generated by shifting the centre of the reconstruction grid along the x and/or y axes within subpixel distance. The same reconstruction algorithm was used for the SR protocol.

Human study

In the human study, SR implementation was applied to reconstruct the high-resolution image. Four human subjects were selected to validate the results of PVE correction. There were three normal subjects and one subject with brain atrophy. All subjects were injected with 185 MBq of ^{18}F -FDG. After the injection, subjects were advised to rest comfortably in a dimly lit room with low noise for 45 min prior to scanning. A 3D brain scan protocol was used to acquire PET data. The total acquisition time was 15 min. The sinogram was reconstructed at five centres of the reconstruction grid to generate five images with different POVs. In these data acquisition phases, a low-resolution image with a matrix size of 336×336 pixels was reconstructed. Consequently, the reconstruction algorithm was performed using the OSEM algorithm with 4 iterations and 24 subsets. The same reconstruction algorithm was used for the SR protocol.

Result analysis

A digital imaging and communications in medicine file of the original PET-reconstructed image was exported to a personal computer for SR reconstruction. Before reconstruction with SR, the red, blue, green (RGB) image was converted into a greyscale image to be compatible with the SR algorithm by ImageJ software (National Institutes of Health, Bethesda, MD). The SR algorithm (see the Super-resolution method section) was then applied. ROIs for each study were drawn to measure mean grey scale value in the ROI as follows: circular ROI using 90% of the diameter from the centre of each sphere for the NEMA/IEC phantom (100% of diameter was not used because there was a pixilated artefact at the edge of the sphere); elliptical ROIs with areas of 150 mm^2 were placed at the caudate and putamen for Hoffman brain phantom and human studies; elliptical ROIs with areas of 200 mm^2 were placed at the medial and lateral temporal grey matter regions for the human study. The greyscale output of the SR image was then converted back into activity concentration (Bq ml^{-1}) with the equation, $y = -264.84x^2 + 710.1x - 222.25$, where x is activity concentration and y is the greyscale value. The formula was generated from the relation between known activity concentration and known greyscale value in the NEMA/IEC phantom ($R^2 = 0.99875$).

Two quantitative values were used to determine the efficiency of the SR algorithm to recover the activity concentration in each sphere size. The first value was the ratio of the measured-to-true activity concentration after PVE correction, the so-called apparent RC (ARC).³ The second value was the percentage yield, which was calculated from the difference between activity concentrations in images reconstructed without and with the use of the SR algorithm divided by that in images reconstructed without the SR algorithm multiplied by 100. Both the ARC and percentage yield were used for analysis in NEMA/IEC phantom and Hoffman brain phantom studies. Only the percentage yield was used in human study, because it is not possible to perform tissue sampling and counting in living humans.

RESULTS

NEMA/IEC phantom study

The PET images of the NEMA/IEC phantom before and after the SR implementation are shown in Figure 1.

It was demonstrated that SR implementation can improve image quality. SR-implemented sphere images had smoother borders

Table 2. Activity concentration, apparent recovery coefficient (ARC) and the percentage yield in the Hoffman brain phantom study

Region	True activity concentration (Bq ml^{-1})	Activity concentration in original image (Bq ml^{-1})	Activity concentration in SR image (Bq ml^{-1})	ARC (original image)	ARC (SR image)	Percentage yield (SR Image)
Left caudate	51,800	37,380	49,220	0.72	0.95	32
Right caudate	51,800	38,850	49,960	0.75	0.96	29
Left putamen	51,800	40,330	45,140	0.78	0.87	12
Right putamen	51,800	32,190	43,290	0.62	0.84	34

SR, super resolution.

Figure 5. The first (a, b), second (c, d) and third (e, f) normal patients' positron emission tomography (PET) brain images in axial anterior commissure–posterior commissure plane. The original PET images are displayed in the left column and the super-resolution images in the right column.

than did originals. The original image looked “pixilated”. Activity concentration after SR image reconstruction was improved in all sphere sizes as seen by ARC in Table 1 and percentage yield in Figure 2. The improvement was substantial in the 10- and 13-cm diameter sphere sizes.

Hoffman brain phantom study

Figure 3 illustrates the PET images of the Hoffman brain phantom reconstructed using the clinical protocol and SR implementation, respectively.

The SR implemented image shows higher intensity and greyscale profile height than does the original image. The resolution of the SR image is also higher than the original by a factor of two. This is evidenced by the image after reconstruction with SR algorithm having a matrix size of 672×672 pixels, whereas the pixel size of the original image has only 336×336 pixels.

Figure 4 shows a zoom-in of a small region of Figure 3 in which the object size is less than two times the FWHM, at which PVE may occur. It is observed that the PVE occurs in the original image as shown by the arrow in Figure 4a and improves in Figure 4b.

Table 2 demonstrates ARC and percentage yield at left caudate, right caudate, left putamen and right putamen from the Hoffman brain phantom study after reconstruction with SR algorithm.

Human study

PET images of the three normal volunteers are shown in Figure 5. Figure 5a,b are the original and SR PET images, respectively, of the first normal subject. There is mild hypometabolism at the anterolateral aspect of the right lateral temporal lobe compared with the contralateral side. The CT scan in this case showed that there was an asymmetrical mild prominent frontal horn of the right lateral ventricle without demonstrable abnormal mass, possibly a normal variation. The

Figure 6. Axial temporal plane positron emission tomography (PET) brain images of the first (a, b), second (c, d) and third (e, f) normal volunteers. The original PET image is displayed in the left column and the super-resolution image in the right column.

Table 3. The percentage yield in different targets of interest in three normal volunteers

Region	Percentage yield		
	First patient	Second patient	Third patient
Right caudate	19.85	23.45	22.62
Left caudate	15.70	23.94	18.69
Right putamen	13.51	18.46	10.95
Left putamen	16.67	18.65	17.17
Left lateral temporal lobe	22.51	27.23	23.91
Right lateral temporal lobe	17.48	22.97	25.43
Left medial temporal lobe	20.43	22.53	25.60
Right medial temporal lobe	21.72	21.15	24.06

mild hypometabolism at the right lateral temporal area in this case may be caused by PVE. Figure 5b shows that the uptake is increased approaching that of the contralateral normal uptake using the SR reconstruction algorithm. The second (Figure 5c) and third (Figure 5e) normal brain scans are those of a 28-year-old female and a 29-year-old female, respectively, and show no hypometabolism. After SR reconstruction (Figure 5d,f), the pixilated artefact in Figure 5c,e disappeared. The overall brain uptake seems to be higher.

Figure 6 shows the image of bilateral temporal lobes in the three normal subjects. After the PVE correction by SR image reconstruction method, the PVE at the lateral and medial temporal areas are apparently corrected.

Table 3 illustrates the percentage yield of caudate, putamen, bilateral lateral temporal areas and medial temporal areas from three normal subjects.

The original and the SR reconstruction axial anterior commissure–posterior commissure plane PET scan of the atrophic brain subject is shown in Figure 7a,b, respectively.

Figure 8 shows the axial temporal plane PET image of the brain atrophy subject. There is symmetrical mild hypometabolism in

the bilateral medial temporal lobes, which may be owing to brain atrophy and resulting PVE.

Percentage yields are shown in Table 4.

DISCUSSION

PET imaging of small source regions such as tumours and certain structures in the brain are subject to PVE-related underestimation of radiotracer concentration (expressed, for example, in terms of the standardized uptake value) and possibly mischaracterization of such structures and misdiagnosis. Most corrections for the PVE require independent estimates of source-region dimensions derived from anatomical imaging modalities.

In this article, we introduce a new technique for brain PET PVE correction by using SR image reconstruction. This method has not been used before for the purpose of PVE correction in brain PET images. The concept of SR technique is that the images are retrieved from different POVs that come from different geometric transformations. In the original SR technique, different POVs are retrieved from different geometric transforms (shifting and/or rotating the object), which is not practical in patient imaging. Information from corresponding pixels in the same region of the low-resolution image is integrated to form a corresponding region in the higher resolution image. The new SR

Figure 7. The axial anterior commissure–posterior commissure plane positron emission tomography image of the atrophic brain subject: (a) original image and (b) super-resolution image.

Figure 8. The axial temporal plane positron emission tomography image of the atrophic brain subject: (a) original image and (b) super-resolution image.

Table 4. Percentage yields in different areas of the normal subject atrophic brain

Region	Percentage yield (%)
Right caudate	30.12
Left caudate	31.43
Right putamen	30.68
Left putamen	32.54
Left lateral temporal lobe	29.12
Right lateral temporal lobe	31.33
Left medial temporal lobe	28.97
Right medial temporal lobe	30.67

implementation by Chang *et al.*²⁰ has an advantage over the original SR implementation in that there is no need for correction from anatomical image and no need to move the patient, which is far more practical in clinical settings. The principle of the new SR technique is that only one raw data image is used to generate different POVs by shifting the centre of the reconstruction grid. The resulting image using the new SR has similar image contrast and resolution as the original SR but achieves slightly lower signal-to-noise ratio (SNR). The new SR implementation has one major advantage of having shorter scan duration and requires the patient to be scanned only once, thus, reducing any effect of patient motion.

The result from the NEMA/IEC phantom study represents the efficiency of SR image reconstruction in terms of improvement of estimating the activity concentration in different sphere sizes. The effect is substantial in smaller sphere sizes demonstrated by greater percentage yield values (Figure 2). Although the SR algorithm substantially reduces the impact of the PVE for small source regions, the correction is not entirely accurate, even for larger spheres (as demonstrated by ARCs being less than one). Thus, additional measures should be incorporated to the SR algorithm to achieve full PVE correction. The effect of SR to improve PVE is also seen in brain phantom studies (Figures 3 and 4, Table 2) and in human brain studies (Figures 5–8, Tables 3 and 4). In human studies, the actual activity concentrations in the brain cannot be measured, thus the ARC cannot be calculated. However, it was shown by percentage yield that all small brain structures showed an increase in activity concentration. The percentage yields in normal volunteers (Table 3) were less than those seen in the atrophic brain subject (Table 4) because the PVE is greater in atrophic and therefore smaller-than-normal brain structures. This may be advantageous in both cognitively impaired patients and patients with temporal lobe epilepsy, since in these patient groups, medial temporal lobe activity concentration is equivocally low. Whether or not the correction affects diagnostic accuracy of the diseased brain has to be proved in further study.

In addition to the PVE correction, the SR algorithm improves subjective image quality in terms of sharpening source region edges by deconvolution of the blurring function (h) and, by upsampling, reduction of pixilation.

The advantages of the SR algorithm over other PVE correction methods are that (1) anatomical information is not necessary, (2) it can be applied to every shape and size of the organ, and (3) it is more practical in clinical settings because it does not increase scan time. The results from the NEMA/IEC phantom, Hoffman brain phantom and human studies demonstrate that SR image reconstruction has the capability to correct the PVE occurring in PET images. SR is suitable for PET study because the initial image of PET is undersampled. The typical PET resolution is about 5–7 mm, which requires <2.5 mm sampling pixel size. However, the initial image that is reconstructed usually has a sampling interval >2.5 mm (otherwise, it will be of high noise), which is undersampled. In this regard, shifting the reconstruction centre will generate different POVs of the same image, which can be combined together to improve overall image resolution. The application of the SR technique to PET image does, however, have a limitation for resolution improvement because it is limited by the system spatial resolution of the PET scanner.

The SR algorithm can reduce the PVE by summation of data in the axial direction. The ARC value after reconstruction with SR algorithm was improved in the NEMA/IEC and Hoffman brain phantoms as shown in Tables 1 and 2, respectively. In the NEMA/IEC phantom, the clinical PET image was acquired with a 256×256 -pixel matrix size. In the Hoffman brain phantom, the clinical PET image was acquired with a 336×336 -pixel matrix size. Owing to the physical limitations of the PET scan itself, clinical protocol to acquire PET data is unable to use any finer matrix size than the two previously mentioned. If the data are acquired in a finer matrix, both the SNR and image quality will suffer. Reconstruction with the SR algorithm yields a finer image matrix, with voxel dimensions approximately one-half of those for images reconstructed using an OSEM algorithm. The finer matrix and resulting smaller voxel dimensions yield reconstructed images with effectively better resolution than coarser-matrix images. Improvement of the PVE from the effect of the smaller pixel size is owing to both the finite resolution effect and tissue fraction effect. The larger the pixel size, the more the activity comes from different types of tissues. As a result, the signal intensity in that particular pixel is the average value of the signal intensity of the underlying tissues. After reducing the pixel dimension, the intensity in each pixel is more accurate.

CONCLUSIONS

The SR algorithm can be used for PVE correction and improvement of image resolution, which is advantageous and practical in clinical brain PET studies.

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