Noise correlation in PET, CT, SPECT and PET/CT data evaluated using autocorrelation function: a phantom study on data, reconstructed using FBP and OSEM

Pasha Razifar$^{1,2}$, Mattias Sandström$^3$, Harald Schneider$^2$, Bengt Långström$^2$, Enn Maripuu$^3$, Ewert Bengtsson$^1$ and Mats Bergström$^{2,4}$

$^1$Uppsala University, Centre for Image Analysis, Lägerhyddsv. 3, SE-752 37 Uppsala.
$^2$Uppsala Imanet AB, Box 967, SE-751 09 Uppsala.
$^3$Uppsala University Hospital, Department of Hospital Physics, SE-751 85 Uppsala.
$^4$Pharmaceutical Biosciences, Uppsala University, Uppsala, Sweden

$^\S$Address correspondence to: Pasha Razifar
Tel: +46-18-66 68 66, Fax: +46-18-66 68 79
E-mail: pasha.razifar@uppsala.imanet.se, pasha@cb.uu.se

E-mail addresses:
PR: pasha.razifar@uppsala.imanet.se, pasha@cb.uu.se
MS: mattias.sandstrom@akademiska.se
HS: harald.schneider@uppsala.imanet.se
BL: bengt.langstrom@uppsala.imanet.se
EM: enn.maripuu@akademiska.se
EB: ewert@cb.uu.se
MB: mats.bergstrom@uppsala.imanet.se
Abstract

Background
Positron Emission Tomography (PET), Computed Tomography (CT), PET/CT and Single Photon Emission Tomography (SPECT) are non-invasive imaging tools used for creating two dimensional (2D) cross section images of three dimensional (3D) objects. PET and SPECT have the potential of providing functional or biochemical information by measuring distribution and kinetics of radiolabelled molecules, whereas CT visualizes X-ray density in tissues in the body. PET/CT provides fused images representing both functional and anatomical information with better precision in localization than PET alone.

Images generated by these types of techniques are generally noisy, thereby impairing the imaging potential and affecting the precision in quantitative values derived from the images. It is crucial to explore and understand the properties of noise in these imaging techniques. Here we used autocorrelation function (ACF) as a technique to study and compare the pattern of noise distributions and correlation in experimentally generated images of PET, CT, PET/CT and SPECT.

Methods
Experiments were performed using phantoms with different shapes. In PET and PET/CT studies, data were acquired in 2D acquisition mode and reconstructed by both analytical filter back projection (FBP) and iterative, ordered subsets expectation maximisation (OSEM) methods. In the PET/CT studies, different magnitudes of X-ray dose in the transmission were employed by using different mA settings for the X-ray tube. In the CT studies, data were acquired using different slice thickness with and without applied dose reduction function and the images were reconstructed by FBP. SPECT studies were performed in 2D, reconstructed using FBP and OSEM, using post 3D filtering. ACF images were generated from the primary images, and profiles across the ACF images were used to describe the noise correlation in different directions. The variance of noise across the images was visualised as images and with profiles across these images.

Results/Conclusions
The results obtained with PET and PET/CT indicate that the pattern of noise correlation is rotation symmetric or isotropic, independent of object shape in images reconstructed using the iterative method. However, noise in FBP images is not isotropic when the shape of phantom is not circular. Also CT images reconstructed using FBP show the same non-isotropic pattern independent of slice thickness and utilization of care dose function. SPECT images show an isotropic correlation of the noise independent of object shape or applied reconstruction algorithm. Noise in PET/CT images was identical independent of the applied X-ray dose in the transmission part (CT), indicating that the noise from transmission with the applied doses does not propagate into the PET images showing that the noise from the emission part is dominant. The results indicate that in human studies it is possible to utilize a low dose in transmission part while maintaining the noise behaviour and the quality of the images.

Index Terms
Positron Emission Tomography, Computed Tomography, PET/CT, Single Photon Emission Tomography, FBP, OSEM, noise, auto-correlation function, isotropic.
Background
Computed tomography (CT) is a technique based on measurement of X-ray transmission through the object to provide visible thin slices through any section in a human; in other words, it is a technique for creating two-dimensional (2D) cross section images of three dimensional (3D) objects [1]. Positron Emission Tomography (PET) and Single Photon Emission Tomography (SPECT) are built on the concept of CT, but use tracing of molecules labelled with positron and gamma ray emitting radionuclides, respectively, to illustrate metabolic and physiological activities in certain organs and tissues. PET/CT combines two state-of-the-art imaging modalities: PET and CT. PET provides high sensitivity functional information of lesions in the body while CT provides detailed information about the location, size, and shape of these lesions, but cannot differentiate pathological lesions from normal structures with the same sensitivity as PET [2-5]. The combined PET/CT scanner has proved to increase the diagnostic value compared to each modality used separately [6].

In PET/CT and SPECT/CT, the CT data are, except for their separate imaging and anatomical delineation, used as transmission data for performing attenuation correction during the reconstruction. PET, SPECT and PET/CT data are usually reconstructed either analytically by filtered back projection (FBP) or iteratively by, for instance, ordered subsets expectation maximisation (OSEM), in SPECT followed by 3D filtering. OSEM needs less computation time compared with earlier versions of iterative reconstructions such as maximum likelihood expectation maximisation (ML-EM) [7]. CT data are reconstructed using FBP with corrections for cone beam geometry and filtered using standard filters related to the imaging task.

In PET and PET/CT the convolution kernel applied on projections in an FBP algorithm is a combination of a ramp filter to cope with blurriness of the image after back projecting the projections [8-9] and a low-pass filter e.g. Hanning filter to damp the high frequency behaviour of ramp filter. FBP is a relatively fast process but it has the drawback that the images are noisier and more sensitive to disturbing factors e.g. patient movements during the scan and, in PET studies, between transmission and emission scans, leading sometimes to loss of information.

In SPECT it has been shown that OSEM gives fewer artefacts compared to FBP with similar uniformity [10-11]. It has also been shown that changing the detector head orbit from circular to elliptical may improve the isotropy of recovered resolution [12].

One limiting factor with respect to the visualisation of discrete changes, e.g. as induced in pathological lesions, is noise in the images. This noise primarily comes from the inherent random variations in the counting of photons and is related to the number of photons detected and used for the generation of images. In CT a high flux of photons is used, giving rise to noise in the images which is about 0.1%, whereas PET and SPECT with fewer photons have noise levels more typically of about 10%.

The main sources of noise in PET images are in decreasing order of magnitude: emission, transmission and blank scans [13]. In tomographic imaging modalities detectors, electronics and recorder systems together create additional noise in two aspects, magnitude and texture [14] where detector system affects only the noise.
magnitude but the recording system affects both magnitude and texture [15]. The choice of reconstruction algorithm and type of convolution kernel used in the reconstruction algorithm significantly affect the magnitude and texture of noise.

Wilson [16] has shown in one of his studies that the magnitude (variance) of noise spreads from regions with higher magnitude in signal towards regions with lower magnitude in signal in FBP reconstruction, but not in iterative reconstruction. Less iteration creates noise, which is correlated at shorter distances compared to FBP algorithm. It has been shown that correlation of each image element influences 1 or 2, pixels of the nearest neighbours in PET [17].

CT has an advantage compared to the two nuclear techniques in its low structural noise in the images. This capacity of low noise and efficient dose use has enabled this technique to visualize low contrast objects [9]. Modern CT equipment has the ability to apply care dose function, with an automated reduction of the dose for non-circular patient cross-sections based on a reduction of the intensity of X-ray tube current at the angular positions at which the patient diameter is smallest. This procedure is performed online with dose regulation during the scanning with preservation of the image quality and noise magnitude [18-21].

Although different aspects of noise have been covered extensively in the literature, we still feel that one aspect has not been adequately covered: the angular dependence of noise correlation in cases when the investigated object is asymmetrical. With asymmetric objects, the count rates will be different in the different acquisition angles and the relative magnitude of the noise is therefore different. It is possible that this angular-dependent noise would in the image reconstruction propagate to the images and there generate a noise correlation which is non-isotropic.

Our present work is based on data from studies on different phantoms with different shapes and sizes, acquired in 2D mode and reconstructed either by FBP or OSEM using different medical imaging modalities. By applying autocorrelation function we explored and compared how these factors affect the properties of noise in images generated with PET, CT, PET/CT and SPECT.

**Methods**

The PET experiments were performed on an ECAT Exact HR+ PET camera (CTI/Siemens, Knoxville, Tennessee) [22]. This unit contains 32 detector rings separated by removable lead septa, and is capable of performing 2D and 3D data acquisition with an axial field of view (FOV) of 15.5 cm. The system generates 63 2D images with a matrix of 128x128 pixels. The CT experiments were performed on a Siemens Somatom Sensation 16 CT scanner (Siemens, Erlangen, Germany). This scanner acquires 16 slices/rotation with a rotation time of 0.5 s. Images are reconstructed into 512x512 matrices; maximum image FOV is 50 cm. The SPECT measurements were made with Millennium VG dual-headed gamma camera with 5/8 “NaI (TI) detectors and a Hawkeye X-ray tomography for attenuation correction (AC) and anatomical information (General Electric Medical Systems, Haifa, Israel). The image matrix was 128x128 pixels. PET/CT experiments were performed employing Discovery ST (D-ST) (GE medical Systems) [23]. The scanner combines a helical multi-slice High Speed Ultra 16 slice, CT scanner and a multi-ring BGO block.
detector PET tomography, which are arranged in 24 rings, spaced by 3.27 mm covering an axialFOV of 15.7 mm. The scanner creates 47 128x128 fused images

In the PET study, the radionuclides $^{11}$C and $^{68}$Ga with 20.3 min and 67.6 min half-lives, respectively, were used. $^{11}$C was produced using a Scanditronix MC-17 cyclotron (Scanditronix AB, Uppsala, Sweden). $^{68}$Ga was obtained from a $^{68}$Ge generator [24]. In the SPECT studies the radionuclide $^{99m}$Tc with 6.02 h was used. $^{99m}$Tc was eluted from a $^{99}$Mo/$^{99m}$Tc generator from Mallinckrodt.

All experiments with each one of the modalities were performed on either a 20-cm-long elliptical torso phantom with 30-cm-diameter long axis and 20-cm short axis or a cylindrical water-filled Nema phantom with 20-cm diameter and 20-cm length [25].

Prior to each PET experiment, a 60-min blank scan was performed with rotating $^{68}$Ge/$^{68}$Ga rod sources without any phantom in the gantry. Then, a 10-min transmission scan was performed where the respective phantom without radioactivity was placed at the centre of the FOV of the scanner. The phantoms were filled with 80 MBq of either $^{18}$F or $^{68}$Ga and 30-min emission scans in 2D mode were made. Finally, to avoid artefacts in the images caused by movement of the object between transmission and emission scan, a 10-min post-injection ('hot') transmission scan was performed. The SPECT emission acquisitions were made with 60x50-s views on phantoms that were filled with 50MBq of $^{99m}$Tc. The used energy window was 140 keV ± 10%. The transmission used for attenuation correction was made with a Hawkeye CT put on half rotation, 140 keV and 3.0 mA. In the CT studies the phantoms were filled with water and the tomography was set to create both two- and three-mm-thick slices. Different numbers of slices were obtained in each scan depending on the desired thickness of slices. Each experiment was additionally performed applying the care dose function. In PET/CT, the study was performed on phantoms filled with 60 MBq $^{68}$Ga in the Nema and 80 MBq $^{68}$Ga in the Torso phantom. The scans were started by transmission (CT) scans using 10 mA, 30 mA and 100 mA setting of the X-ray tube followed by 12x30-min emissions.

The PET images reconstructed using the initial transmission scan for attenuation correction were not used because of slight observed positioning errors when replacing the phantom after filling it with radioactivity. Instead the transmission made after the emission was used for attenuation correction in the reconstruction process, including the segmentation technique, as included in the ECAT 7.2 software (CTI, Knoxville, Tennessee).

In PET and PET/CT, both FBP and attenuation-weighted OSEM as included in the scanner software were used for reconstructing the images. Different types of filters can be used with each of the reconstruction methods. A low pass Hanning filter with 4 mm (FWHM) was used in this study. The use of the same filter with FBP and OSEM ensures similar spatial resolution. PET/CT images were reconstructed with three different combinations of CT transmission data (10, 30 and 100 mA) and emission data. All CT images were reconstructed using FBP with corrections for cone beam geometry and filtered using standard abdomen filter, as included in the scanner software. All SPECT images were reconstructed using both FBP and OSEM algorithms, as included in the Entegra workstation, using the default settings for filters and number of iterations in OSEM (Hanning filter with a cut-off of 0.85 and 8 subsets
and 4 iterations). Iterative Reconstruction Attenuation Corrected with applied 3D post filtering (IRACF) was used as the nomenclature for iteratively reconstructed SPECT data in this study.

A programme was developed to calculate the ACF of the reconstructed PET images, performed both in frequency and spatial domain. The spatial equation is based on 2D cross-correlation of the matrix:

\[
C_{orr}(k,l) = \sum_{i=1}^{m} \sum_{j=1}^{n} a_{i,j} a_{i+k,j+l}
\]  

(1)

where \(k\) and \(l\) refer to lags of the function and

\[
\max(1,1-k) \leq i \leq \min(m,m-k)
\]

and

\[
\max(1,1-l) \leq j \leq \min(n,n-l)
\]

To avoid disturbing effects at the edges of the FOV an arbitrary central slice within each frame was used. Subsequently, an arbitrary chosen matrix with a size of 25x25 from the central part of the slice was used as a mask when applying the ACF. After subtraction of the average over this matrix, the ACF image was generated showing the correlation of the noise in the image. The resulting image was then normalized by dividing each pixel value by the maximum pixel value within the generated ACF image. The results from this procedure applied to images from all experiments were studied and compared.

The aim of the ACF application was to describe the noise correlation between the pixels within each image. A specific aim was to analyse the form and the shape of the 2D autocorrelation function in the images in different imaging modalities. The programme results in images, which can be used for the visualization and comparison of images using different techniques obtained with different reconstruction algorithms and for plotting 1D vertical and horizontal profiles through the images.

Another programme was developed to calculate the variance matrix as indicator of the noise magnitude distribution across the image plane. The method compares pixel values through several adjacent slices within a frame and calculates

\[
S_j^2 = \sum_{i=1}^{n} (x_{ij} - \overline{x}_j)^2 / (n-1)
\]  

(5)

where \(n\) is the total number of slices and \(I\) refers to each pixel in each image vector and \(j\) is equal to number of column times number of rows in image matrices. E.g., in PET, \(j\) is equal to 128x128. The aim of this application was to study the magnitude of noise in relation to position within the images produced by different modalities.
results are illustrated and studied as 2D variance images and 1D horizontal profiles across the images.

**Results**

**PET studies**

In the study on the NEMA phantom (Fig. 1), the results indicate an identical and isotropic ACF with a similar pattern of noise texture independent of applied reconstruction methodology and used filter (6 mm Gaussian and 4 mm Hanning produced identical results). In the torso phantom (Fig. 2), however, the results of the ACF indicate a non-isotropic correlation of the noise in images reconstructed using FBP, independent of used filter (6 mm Gaussian and 4 mm Hanning). On the other hand, the images reconstructed using OSEM (Fig. 3), show an identical and isotropic form with a similar pattern of noise texture independent of used filter.

(Figure 1)

(Figure 2)

(Figure 3)

**CT Studies**

In CT images of the circular Nema phantom, the ACF shows the expected isotropic behaviour. When the elliptic torso phantom was scanned, a slightly non-isotropic behaviour was observed as indicated by the ACF (Fig. 4 -5). This non-isotropy is only slightly affected by the use of the care dose. Identical results were obtained with 2- and 3-mm slice thickness.

(Figure 4)

(Figure 5)

**SPECT studies**

In the studies of noise autocorrelation in SPECT images, the images from the cylindrical phantom showed close to but not entirely isotropic behaviour. Images from the elliptic phantom showed an ACF which was close to isotropic (Fig. 6). The same was true when the images were reconstructed with the IRACF iterative reconstruction.

(Figure 6)

**PET/CT studies**

The CT images (Fig. 7) and the PET images (Fig. 8) from the cylindrical phantom using PET/CT showed an expected isotropic behaviour, which in the case of PET was independent of reconstruction method. For scanning and reconstruction images from the elliptic phantom; however, both the CT images (Fig. 9) and PET images (Fig. 10) showed a slightly non-isotropic behaviour. When the PET image was reconstructed
using OSEM, the noise correlation became more isotropic (Fig. 11). There was an identical noise correlation pattern in the PET images reconstructed with different CT doses.

(Figure 7)
(Figure 8)
(Figure 9)
(Figure 10)
(Figure 11)

**Variance images**

**PET studies**
In the PET images obtained from the cylindrical and elliptic phantom, the variance of the noise is non-uniform across the image field. This is apparent under all conditions (Fig. 12-13) but with slightly different features depending on reconstruction algorithm. With FBP a high noise variance gradually decreases from inside to outside of the phantom, whereas with OSEM, the noise decreases abruptly at the border of the object.

(Figure 12)
(Figure 13)

**CT studies**
The variance across the CT image field shows a peak at the centre of the object and gradually decreases towards the periphery. The same feature is observed in the cylindrical and the elliptic phantom (Fig. 14). The variance at the centre is about three times higher than that at the periphery. similar behavior is observed independently of slice thickness and application of care dose.

(Figure 14)

**SPECT studies**
In the SPECT images, the noise magnitude across the image field shows a different behaviour as compared to PET. The noise is more evenly distributed although in FBP images the noise variance distribution is more global compared with images reconstructed by IRACF (Fig. 15-16).

(Figure 15)
(Figure 16)
PET/CT studies

Fig. 17 visualizes the noise distribution over the image field for the CT in PET/CT. The noise is significantly higher in the centre of the cylindrical phantom, similar to that observed in the previous study with CT. With the cylindrical phantom the variation along the horizontal axis is less than along the vertical axis. Almost identical results were obtained using different magnitude of X-ray dose.

(Figure 17)

Sinogram data in PET and SPECT

To understand the noise contribution, profiles were generated across the sinograms from PET studies with the two phantoms. The counts recorded from the circular phantom dipped slightly in the central part (Fig. 20). For the elliptic phantom this central dip was more accentuated in the plot over the short axis representation but the profile was rather flat in its central portion for the long axis representation.

The corresponding plots for SPECT showed a different behaviour, with higher counts centrally and rather similar count values for the two projections in the elliptic phantom (Fig. 21).

(Figure 20)

(Figure 21)

Discussion

The aim of the present study was to explore and compare the properties of noise, notably its correlation, in images from CT, PET, PET/CT and SPECT. We believed that despite their similarity as tomographic imaging devices, the fact that the acquisition of detector data and modes of reconstruction differ could lead to differences in the expression of noise in the images. CT is based on measurement of transmitted X-ray photons from X-ray tube through the object and to the detectors. PET is based on the simultaneous measurement of two annihilation photons that emerge from the body and hit each of two detectors on opposite sides of the object. SPECT is based on recording single photons that emerge from the body and acquisition of a large number of projections by sequential rotation of the detector system around the body. For these techniques the measured data, typically organised in projections are subjected to different corrections. For PET and SPECT dominated by attenuation correction and the projection data are then utilised in a reconstruction algorithm for the generation of images. The reconstruction part can be similar for the three devices with the predominant methods being FBP or iterative reconstruction.
Since noise in the images is a factor important for the visualisation of discrete signals and the generation of quantitative values, it is important to understand its features. To illustrate its features we developed a programme to generate auto-correlation images which predominantly indicate the noise correlation and the angular dependence of noise correlation. Images depicting the variance across the image planes were created to indicate the spatial dependence of noise. Some of the characters of noise were explained by studying the profiles of counts in different directions in the sinogram domain in PET and SPECT.

Applying the auto-correlation function on the reconstructed PET, PET/CT and SPECT data revealed a clear correlation: the noise was affected by the adjacent 2-3 pixels. This correlation was similar for all devices and all reconstruction algorithms. An expected, an isotropic pattern of correlation of the noise in the images was obtained in the circular Nema phantom. In CT scans from both regular CT and PET/CT, the same behaviour was observed.

However, the noise correlation pattern obtained from the torso phantom with elliptical shape was clearly non-isotropic when examined with PET or PET/CT and dissimilar between images using the two reconstruction methods. When studies were performed on this non-circular phantom, ACF images from PET, CT and PET/CT studies reconstructed using FBP showed an asymmetric correlation texture of the noise. This non-isotropic behaviour can be explained in three different ways; the first explanation is that the relative noise in sinogram data in the study on Torso phantom differs in different directions. The sinogram data indicate that the count rates are higher in the direction of the shortest diameter compared to the angle with the longest diameter. This is because along the 30 cm axis the detectors see a 20 to 30 times higher amount of radioactivity. Still, this radioactivity is in turn subjected to an attenuating path that is 10 cm longer, leading to an overall 43% lower count rate. Since noise in the projection data is Poisson distributed, (Omit?this means that) the noise standard deviation is angular dependent and 25% higher along the long axis of the phantom. The second explanation is related to the attenuation correction whereby the projection data with its noise is multiplied by attenuation correction factors which are different for different projections. The third contribution is related to the way noise is handled in the FBP algorithm when the noise magnitude differs in the different projections. The noise modification and correlation induced by the convolution filter in the reconstruction is the same in all projections. The back projection then distributes noise with different magnitude in different angular directions.

The correlation pattern of the noise in images from PET, CT and PET/CT studies become close to symmetric when the data are reconstructed using OSEM. This behaviour depends on how the iterative reconstruction algorithm handles the noise with an inherent attempt to iterate to similar deviations for each angular projection, which then tends to equalise the noise magnitude for the different projections.

In PET/CT studies the results from fused images were identical independent of the applied dose of X-ray in transmission part (CT). This observation indicates that the noise from the transmission at these doses does not propagate into the fused images because the noise from the emission part is dominant. The result suggests that in human studies it might be possible to use a low dose in the CT transmission part while
maintaining the noise behaviour and the quality of the images. This observation could lead to lower X-ray dose to the patient.

In SPECT studies, the correlation pattern was still isotropic in images obtained from an elliptical object and independent of the reconstruction methodology used. This behaviour could be because the number of counts in different directions in the study on either circular or elliptical phantom is almost the same. This similarity is caused by the way radioactivity is sensed, with a detector which records the superficial activity deep inside the object. Since attenuation is a dominant factor with respect to recording of radioactivity, the number of counts will be come relatively similar for a uniform elliptic phantom. On the other hand, a focal radioactivity not centred in the phantom will instead give highly variable counts in different detectors. Additionally the detector sensitivity is significantly affected by distance from the detector. This could explain a slightly non-isotropic effect observed in the circular phantom that was not perfectly centred.

ACF images from CT studies reconstructed by FBP on data from the Torso phantom showed a non-isotropic form with a similar pattern of noise texture independent of thickness of the sliced images. The correlation pattern was broader along the horizontal axis compared with the vertical. This non-isotropic behaviour depends on how data is acquired in CT. The X-ray tube rotating around the object emits X-rays, which are detected by detectors that are placed as a block on the opposite side of the tube. The number of photons detected is highly dependent upon the thickness of the object, and the relative noise is hence larger in the horizontal direction of the object. These differences in noise magnitude are, as indicated above, handled by the FBP algorithm such that a non-isotropic noise correlation is given in the images.

The results from studies on CT images with applied care dose function gave slightly different noise correlation compared to those without care dose. Yet the noise correlation was not identical in the vertical and horizontal direction, suggesting that the care dose could not fully equalise the relative noise in different directions.

The variance across the PET images shows a significantly broader distribution with FBP than with OSEM. With OSEM, the variance reduces abruptly at the border of the object. The variance is significantly larger centrally in the object than at the periphery.

The combined effect of noise correlation for asymmetric objects and a varying noise variance across the image field significantly complicates, the interpretation of the images when statistical methods are used, such as with statistical estimates of precision in average values, use of statistical parametric mapping methods and principal component analysis. Hence it is recommended that iterative reconstruction methods are used for such applications.

**Competing interests**

The authors declare that they have no competing interests.

**Authors' contributions**
Authors PR and MB designed the study. They created the method for applying ACF, performed the image and data analysis and drafted the paper.

Author MS helped perform the SPECT studies, acquire and reconstruct the data and write this paper.

Authors HS, EB, EM and BL helped with some practical approaches and the writing of this paper.

**Acknowledgements**

The authors wish to thank the staff at Uppsala Imanet AB, especially Mr. Lars Lindsjö for his assistance with the measurements, and the staff of the chemistry department for the radionuclide production. Our thanks to Uppsala University Hospital, Departments of Hospital Physics and Radiology for their assistance and help.

**References**


Figure Captions

Figure 1 - Results of the cylindrical NEMA phantom study. Acquired image (upper left) reconstructed using FBP with applied 6 mm (FWHM) Gaussian filter, corresponding ACF image (upper right) and vertical (dash point) and horizontal profiles (dash star) through the ACF image.

Figure 2 - Results of the elliptical torso phantom study. Image (upper left) reconstructed using FBP with applied 6 mm (FWHM) Gaussian filter, corresponding ACF image (upper right) and vertical (dash point) and horizontal profiles (dash star) through the ACF image.

Figure 3 - Results of the elliptical torso phantom study. Image (upper left) reconstructed using OSEM with applied 6 mm (FWHM) Gaussian filter, corresponding ACF image (upper right) and vertical (dash point) and horizontal profiles (dash star) through the ACF image.

Figure 4 - Results of the cylindrical Torso phantom in CT study. FBP image with thickness of 2mm scanned with constant mA setting (upper left) respective with applied care dose (upper right). ACF images from the normal scan (lower left) respective with applied care dose (lower right).

Figure 5 - 1D vertical and horizontal profiles through the ACF image based on the normal scan (upper) respective with applied care dose (lower).

Figure 6 – Results of SPECT study on elliptical Torso phantom. Image reconstructed using FBP followed by 3D filtering (upper left) and corresponding ACF (upper right) followed by 1D vertical and horizontal profiles of the ACF (down).

Figure 7 – Results of PET/CT study on the cylindrical Nema phantom. The CT image reconstructed using FBP (upper left) and corresponding ACF (upper right) followed by 1D vertical and horizontal profiles of the ACF (down).

Figure 8 – Results of PET/CT study on the cylindrical Nema phantom. The PET image reconstructed using FBP with applied 4-mm Hanning filter (upper left) and corresponding ACF (upper right) followed by 1D vertical and horizontal profiles of the ACF (down).

Figure 9 – Results of PET/CT study on the elliptical Torso phantom. The CT image reconstructed using FBP (upper left) and corresponding ACF (upper right) followed by 1D vertical and horizontal profiles of the ACF (down).

Figure 10 – Results of PET/CT study on elliptical Torso phantom. PET image reconstructed using FBP with applied 4-mm Hanning filter (upper left) and corresponding ACF (upper right) followed by 1D vertical and horizontal profiles of the ACF (down).
Figure 11 - Results of PET/CT study on elliptical Torso phantom. PET image reconstructed using OSEM with applied 4-mm Hanning filter (upper left) and corresponding ACF (upper right). 1D vertical and horizontal profiles of the ACF (down).

Figure 12 - Noise variance across the image field for the cylindrical NEMA phantom reconstructed using FBP (upper left) and OSEM (upper right). 1D horizontal profile through the variance images of FBP (lower left) OSEM (lower right) reconstructed images.

Figure 13 - Variance images from the PET study of the Torso phantom, reconstructed using FBP (upper left) and OSEM (upper right). 1D horizontal profiles through the respective images (lower left and right).

Figure 14 - Variance across the CT image of the cylindrical (left) and elliptic (right) phantom. Horizontal profiles over the corresponding variance images are shown in the graphs.

Figure 15 - Results of the SPECT study on the cylindrical NEMA phantom. Variance image reconstructed using FBP (upper left) and IRACF (upper right). 1D horizontal profile through the variance image reconstructed using FBP (lower left) and reconstructed using IRACF (lower right).

Figure 16 - Results of the SPECT study on elliptical Torso phantom. Variance image reconstructed using FBP (upper left) and IRACF (upper right). 1D horizontal profile through the variance image reconstructed using FBP (lower left) and reconstructed using IRACF (lower right).

Figure 17 - Results of the PET/CT study on cylindrical NEMA phantom. Variance images of CT (transmission scan) of Nema (upper left) and Torso phantom (upper right). Corresponding 1D horizontal profile through the variance images of Nema (lower left) and Torso phantom (lower right).

Figure 18 – PET results of PET/CT study on cylindrical NEMA phantom. Variance images reconstructed using FBP (upper left) and OSEM (upper right). 1D horizontal profile through the variance image reconstructed using FBP (lower left) and reconstructed using OSEM (lower right).

Figure 19 - Results of PET/CT study on elliptical Torso phantom. PET variance images reconstructed using FBP (upper left) and OSEM (upper right). 1D horizontal profile through the variance image reconstructed using FBP (lower left) and reconstructed using OSEM (lower right).

Figure 20 - Sinograms from PET study of Nema (left) and Torso (right) phantom illustrating detector counts for the different projections (upper images). Profiles through the end and central parts of the sinograms, corresponding to horizontal vs. diagonal projections during acquisition (lower plots).
Figure 21 - Projections obtained at two perpendicular angles at the acquisition with SPECT for the Torso phantom. Profiles through the two different projections, corresponding to horizontal vs. diagonal direction in SPECT acquisition (down). The position shift is related to the fact that the phantom was not centred at the rotation axis.
Figures

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Figure 15: Variance image of Nema FBP-Slice 55-85 and Variance image of Nema IRACF-Slice 55-85.

1D horizontal profile FBP-Slice 55-85 and 1D horizontal profile IRACF-Slice 55-85.
Figure 19

Variance image of 30mA-PET-CT-Toso-FBP

1D horizontal profile of 30mA-PET-CT-Toso-FBP
Additional files provided with this submission:

Additional file 1: CoverLetterToBMCMedicalImaging_DiffMod_article_20050326.doc : 31KB
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