1. **Background**

Due to the significant improvement provided in image quality, iterative reconstruction methods such as OSEM (Ordered Subsets Expectation Maximization) have now generally replaced traditional analytic methods such as filtered backprojection for PET imaging. Since iterative methods were first introduced in commercial systems, advancements in computing power and algorithm design have resulted in more accurate models being utilized in the reconstruction algorithms, providing further improvements in image quality. This includes improvements in components of the system model such as accurate projectors based on the detector geometry [1], fully 3D scatter estimates [2], and low noise randoms estimates [3]. In addition, the model used in the iterative algorithm has been improved to more accurately incorporate the statistics of the measured data by including all corrections in the iterative update equation [1]. This paper describes a further improvement to the system model to include the measured detector response of the PET scanner in the iterative system model [4-5].

2. **SharpIR basics**

The purpose of the SharpIR algorithm is to improve PET image contrast to noise by incorporating information about the PET detector response into the 3D iterative reconstruction algorithm. The SharpIR algorithm is a feature which can be used with the VUE Point HD (VPHD, Fully 3D Iterative Reconstruction) and VUE Point FX (VPFX, Fully 3D Time-of-Flight Iterative Reconstruction) algorithms. The detector response is a function of several scanner parameters including detector sampling width, detector geometry, and parallax effects, all of which contribute to a spatially variant blurring point response. The detector response of a scanner can be characterized with Monte Carlo simulations [4] or experimental measurements [5]. The incorporation of the detector response into the 3D iterative reconstruction algorithm can be included in either the image domain [6], or the sinogram domain [5]. The SharpIR algorithm follows the methodology of A. Alessio et al. [5] by including detector response information in sinogram space. The detector response of the scanner can be measured by placing a point source within the scanner and acquiring scan data at varying locations in both the radial and axial dimension as shown in **Figure 1**. Because of the tight mechanical tolerances of Discovery® PET/CT 600 series scanners, a common detector response function can used for each scanner type.

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**Figure 3** shows normalized axial sinogram profiles from point source acquisitions on the Discovery PET/CT 600. As expected less variability is seen with axial position than with radial position. The radial and axial detector response measurements are taken at discrete locations across the gantry. These fixed measurements are then mapped to a model for each sinogram element of the scanner. Once the detector response is modeled over the entire detector space, a detector response matrix is created which can be incorporated into the VUE Point HD and VUE Point FX algorithms. To simplify the implementation, the detector response matrix can be separated into radial and axial components.

The OSEM equation can be written as:

\[
I_{i+1} = \sum_m I_m \sum_n H_{m,n} M_{m,n} P_{m,n} + A_{m,n}
\]

**Equation 1**

where:

- \( I_i \) represents the \( n \)th element of the \( i \)th update of image \( I \),
- \( P_m \) represents the \( m \)th element of the measured emission sinogram,
- \( M_m \) represents the \( m \)th element of the estimated "multiplicative" matrix. This includes normalization, deadtime, and attenuation effects.
- \( A_m \) represents the \( m \)th element of the estimated "additive" matrix. This includes scatter, randoms, and keyhole (for targeting) sinograms.
- \( H_{m,n} \) represents the geometry of the system.

Incorporating the detector response function into **Equation (1)** results in a modified version of the update equation where \( H \) is replaced by \( H' \). \( H' \) now includes the system response as:

\[
H'_{m,n} = \sum_k D_{m,k} H_{k,n}
\]

**Equation 2**

where \( D \) includes both the radial and axial detector response model. While this equation expresses the detector response as a general function of all projection data, in practice the model is implemented as a convolution along the rows of the sinogram and a convolution along the axial direction in the projection planes.

### 3. Results

#### 3.1 Phantom Results

**Resolution**

A capillary tube was filled with FDG and scanned at varying distances from the isocenter on both the Discovery PET/CT 600 and the Discovery PET/CT 690. On the Discovery PET/CT 600, images were reconstructed with VUE Point HD and VUE Point HD-SharpIR. On the Discovery PET/CT 690, images were reconstructed with VUE Point FX and VUE Point FX-SharpIR. The VUE Point HD/FX images were reconstructed with 4 iterations, a pixel size of 0.98 mm, and a 2 mm Full Width Half Max (FWHM) Gaussian in-plane post filter. The VUE Point HD/FX-SharpIR images were reconstructed with 10 iterations, a pixel size of 0.65 mm, and no post filter. Twenty four subsets were used on the Discovery PET/CT 690, 32 subsets were used on the Discovery PET/CT 600.  FWHM measurements were obtained by fitting the three central points with a parabola and using linear interpolation to determine the half maximum points. Results shown in **Figure 4** are the average of radial and tangential FWHM measurements. The results demonstrate an improvement in resolution with the SharpIR algorithm for a line source in air. Note that as described in A. Alessio et al. [5], a line source measured in a more realistic imaging scenario such as a warm background will have reduced resolution improvement with SharpIR.

**Figure 4**: FWHM resolution measurements of a line source acquired on the (a) Discovery PET/CT 600 and (b) Discovery PET/CT 690. The VUE Point HD/FX images were reconstructed with 4 iterations, a pixel size of 0.98 mm, and a 2 mm FWHM Gaussian in-plane post filter. The VUE Point HD/FX-SharpIR images were reconstructed with 10 iterations, a pixel size of 0.65 mm, and no post filter. Twenty four subsets were used on the Discovery PET/CT 690, 32 subsets were used on the Discovery PET/CT 600.
**Contrast Recovery**

A NEMA Image Quality phantom (PTW, Freiburg, Germany) was scanned on the Discovery PET/CT 690. Images were reconstructed with both VUE Point FX and VUE Point FX-SharpIR with 1.56 mm pixels and a 2 mm FWHM Gaussian in-plane post filter. Reconstructed images are shown in Figure 5. The VUE Point FX images shown were reconstructed with 4 iterations and 24 subsets, and the VUE Point FX-SharpIR images were reconstructed with 6 iterations and 24 subsets. Contrast vs background variability plots as a function of iteration number (1-10) are shown in Figure 6. The plots demonstrate improved contrast recovery in the NEMA Image Quality Phantom with SharpIR. The VUE Point FX-SharpIR algorithm requires more iterations to reach convergence than VUE Point FX. This is not unexpected because VUE Point FX-SharpIR has a system model which relates each voxel to more measurement locations than does VUE Point FX, thus incorporating more information into the reconstruction algorithm.

![Figure 5: NEMA Image Quality Phantom images reconstructed on the Discovery PET/CT 690. Images were reconstructed with VUE Point FX (left) and VUE Point FX-SharpIR (right). The VUE Point FX images shown were reconstructed with 4 iterations and 24 subsets, and the VUE Point FX-SharpIR images were reconstructed with 6 iterations and 24 subsets. The pixel size was 1.56 mm and a 2 mm FWHM Gaussian in-plane post filter was applied.](image)

To demonstrate the effect of SharpIR on more complex objects, a Hot Spot Insert™ (Data Spectrum, Hillsborough, NC) centered in the FOV was scanned on the Discovery PET/CT 690. The phantom consists of a set of hot rods in a cold background, along with a uniform water section. A transaxial image of the hot rod section of the phantom reconstructed with VUE Point FX and VUE Point FX-SharpIR is shown in Figure 7. A set of ROIs was drawn on the CT images over both the 4.8 mm rods and the 6.4 mm rods. These ROIs were then transferred to the PET images where activity values were computed for both sets of rods (4.8 mm and 6.4 mm). As a measure of contrast recovery, the average activity in the rods was compared to the activity of a large ROI drawn in the uniform water section of the phantom. Images were reconstructed with VUE Point FX and VUE Point FX-SharpIR for 1-50 iterations with 24 subsets, a pixel size of 1.95 mm and a 2 mm FWHM Gaussian in-plane post filter. Contrast recovery was plotted vs noise for each iteration, where noise was computed as the standard deviation of a large ROI in the water section. Results are shown in Figure 8 which demonstrate that incorporating detector response in the reconstruction improves contrast recovery. The results also demonstrate slower convergence for the small rods.

![Figure 7: Transaxial images of the Hot Spot Insert Phantom (Data Spectrum) acquired on the Discovery PET/CT 690. The images were reconstructed with VUE Point FX (left) and VUE Point FX-SharpIR (right). Images were reconstructed with a pixel size of 1.95 mm and a 2 mm FWHM Gaussian in-plane post filter. Images shown were reconstructed with 15 iterations and 24 subsets.](image)

![Figure 6: Percent contrast recovery of 10 mm NEMA Image Quality Phantom sphere acquired on the Discovery PET/CT 690. Images were reconstructed with VUE Point FX (blue) and VUE Point FX-SharpIR (magenta). Note the results are averages of three NEMA acquisitions per NEMA 2007 recommendations.](image)
4. Clinical Results

4.1 Whole Body Imaging

A whole body FDG scan acquired on the Discovery PET/CT 690 was reconstructed with VUE Point FX and VUE Point FX-SharpIR. Reconstructed images are shown in Figure 9. The VUE Point FX image was reconstructed using 2 iterations/24 subsets and a post-filtering of a 4 mm FWHM Gaussian in-plane and a [1 6 1] weighted axial filter (“Light” axial filter). The VUE Point FX-SharpIR image used the same parameters except that 3 iterations were performed. For both images, the pixel size is 2.73 mm.

To demonstrate contrast recovery, an ROI was drawn on a metabolic hot spot and also on a uniform region of the liver. A plot of contrast (metabolic hot spot activity/liver activity) vs liver noise as a function of iteration number (1-10) is shown in Figure 10. The results of Figure 10 demonstrate an improvement in contrast to noise with SharpIR.

Figure 8: Contrast recovery of (a) 4.8 mm spheres and (b) 6.4 mm spheres in the Hot Spot Insert. Contrast recovery is computed as the ratio of average activity in the hot rods divided by the average activity in a large ROI drawn in the water section. Noise is computed as the standard deviation of the water section ROI. Contrast recovery is shown for VUE Point FX (blue), VUE Point FX-SharpIR (magenta).

Figure 9: Clinical FDG images acquired on the Discovery PET/CT 690. Images reconstructed with VUE Point FX (left) and VUE Point FX-SharpIR (right). The VUE Point FX image was reconstructed using 2 iterations/24 subsets and a post-filtering of a 4 mm FWHM Gaussian in-plane and a [1 6 1] weighted axial filter (“Light” axial filter). The VUE Point FX-SharpIR image used the same parameters except that 3 iterations were performed. Pixel size for both images was 2.73 mm. ROIs were drawn on the metabolic hot spot shown with the arrow and also on the liver. ROIs were used for plots of Figure 10.

Figure 10: Contrast (metabolic hot spot activity/liver activity) plotted vs liver noise for iterations 1-10. Images reconstructed from Discovery PET/CT 690 whole body FDG scan. Plots are shown for VUE Point FX and VUE Point FX-SharpIR using an ROI from the metabolic hot spot shown in Figure 9 as well as an ROI in the liver. The vertical axis shows contrast computed as the metabolic hot spot ROI values divided by the average value of the liver ROI. The horizontal axis is liver noise computed as the standard deviation in the liver.

A whole body FDG scan acquired on the Discovery PET/CT 600 was reconstructed with VUE Point HD and VUE Point HD-SharpIR. Reconstructed images are shown in Figure 11. The VUE Point HD image was reconstructed with 2 iterations/32 subsets and post-filtering of a 6.4 mm FWHM Gaussian in-plane and a [1 4 1] weighted axial filter applied (“Standard” axial filter). The VUE Point HD-SharpIR image was reconstructed with 3 iterations/32 subsets with post-filtering of a 4 mm
FWHM Gaussian in-plane and a [1 6 1] weighted axial filter ("Light" axial filter). For both images, the pixel size is 2.73 mm. To demonstrate contrast recovery, an ROI was drawn on a metabolic hot spot and also on a uniform region of the liver. A plot of contrast (metabolic hot spot activity/liver activity) vs liver noise as a function of iteration number (1-10) is shown in Figure 12. The results of Figure 12 demonstrate an improvement in contrast to noise with SharpIR.

4.2 Brain Imaging

A clinical brain scan acquired on the Discovery PET/CT 690 was reconstructed with VUE Point HD and VUE Point HD-SharpIR. Reconstructed images are shown in Figure 13. The VUE Point HD images were reconstructed with 4 iterations/24 subsets, the VUE Point HD-SharpIR images were reconstructed with 8 iterations/24 subsets. Pixel size was 1.56 mm with a 3 mm FWHM Gaussian in-plane post filter was applied. The images demonstrate a qualitative increase in contrast vs noise with the SharpIR reconstructions.

5. Summary

By incorporating more information about the detector system response, the SharpIR algorithm improves the accuracy of underlying model used in both VUE Point HD and VUE Point FX. Because SharpIR has a system model which relates each voxel to more measurement locations than VUE Point HD or VUE Point FX, it results in a more sophisticated reconstruction model which may take more iterations to solve [5]. Like any change in reconstruction processing, the effects of the SharpIR algorithm should be considered when evaluating longitudinal studies. If desired, SharpIR can be turned off with both VUE Point HD and VUE Point FX, providing a method for a more consistent interpretation across longitudinal studies. The results presented here demonstrate that the SharpIR algorithm improves spatial resolution for a line source in air, and improves contrast recovery at equivalent noise levels in phantoms and clinical studies.
References


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