PHYSICS, ORIGINAL ARTICLE

Evaluation of the Image Contrast with a New Depth-Dependent Collimator Resolution Iterative Reconstruction Method

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ABSTRACT

Iterative reconstruction methods are well suited to improve image quality and clinical performance in SPECT imaging in nuclear medicine. This is done by incorporating internal modeling of the imaging physics in SPECT reconstructions to correct for the major factors affecting image quality. These factors include degradation of spatial resolution with increasing distance of the source from the collimator, Compton scatter and photon attenuation. Since noise suppression approaches degrade the image resolution, a good balance of resolution recovery and noise suppression is desirable. The Astonish software package provides powerful control of the resolution recovery technique and an application-specific optimization of noise suppression. ***Objectives:*** The aim of the study is to compare the contrast value of the images resulting from filtering with different available iteration methods with the new depth dependant collimator resolution iteration technique (Astonish) in SPECT imaging. ***Materials and methods:*** The study was performed on a Forte dual head gamma camera, using the SPECT phantom. The contrast value is calculated for a specified hot sphere within the cold background by applying the studied reconstruction iterative methods (MLEM, 3D-OSEM and Astonish) with iteration number from 1-15 iterations. The resultant calculated contrast values are then compared for each iteration method. ***Results***: Regarding the **MLEM** iteration method the average contrast value was [0.8002 ± 0.0304] and for the **3D OSEM** iteration method the average contrast value was [0.8662 ± 0.0068]. As regards to the **Astonish** iteration method the average contrast value was [0.907 ± 0.0394]. There was a statistical significant difference in contrast value between the studied iteration methods (p=4.2E-12).

Conclusion: The MLEM iteration method increases the sphere count with increasing the iteration number; and also it increases the background counts with about 3 fold that of the 3D OSEM method, and about 4 folds that of Astonish method. Although both 3D-OSEM and Astonish methods give sphere counts that are almost within the same range but incorporating the depth dependant resolution recovery parameter in the Astonish method leads to decrease in the background count with increasing the iteration number, which in turn enhances the contrast of the resultant image from Astonish iteration method rather than that of both MLEM and 3D-OSEM methods.

Key words: Astonish, Iteration, Contrast

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INTRODUCTION

Nuclear medicine SPECT images suffer from noise due to low count statistics and poor resolution. These effects may limit the clinical applications of nuclear medicine procedures. Numerous methods have been proposed to compensate for these effects, with the goal of improving the quality of the final reconstructed image. Noise in SPECT images is introduced by the lack of detected photons. In general, the limited dose of radiopharmaceutical introduced to the patients, the limited patient scanning time, patient attenuation, the decreased sensitivity due to the use of collimators, as well as the intrinsic detection efficiency of the SPECT cameras limit the number of detected photons in the order of 10^3-10^4 counts per second and 10^6-10^7 per patient scan [(1)].

The resolution of a SPECT system is determined by the intrinsic resolution of the camera (mainly the crystal) and the geometric resolution of the collimator. The geometric resolution depends upon the collimator hole-size and hole-length and also depends upon the distance the object is from the collimator surface. The far the object is from the collimator surface, the poorer the resolution. The overall system resolution gets poorer with increased distance from the object to the detector surface (denoted as depth-dependent here after). It is usually determined by measuring the full-width at half-maximum (FWHM) of point sources imaged at different distances from the collimator surface [(2)].

Iterative reconstruction methods are used to improve image quality and clinical performance. This is done by incorporating internal modeling of the imaging physics in SPECT reconstructions to correct for the major factors affecting image quality. These factors include degradation of spatial resolution with increasing distance of the source from the collimator, compton scatter and photon attenuation. The degradation of spatial resolution with distance from the collimator reduces the overall spatial resolution in the images and can produce non-uniformity artifacts in the investigated organ. Scattered photons detected in the energy window reduce image contrast and interfere with the ability to efficiently perform attenuation correction. Non-uniform photon attenuation may produce artifacts that can mimic perfusion defects if not corrected [(3)]. Given the fact that SPECT is a count-starved imaging modality, three major approaches have been investigated by the nuclear medicine community to suppress the noise in SPECT images in addition to the tremendous effort of improving SPECT system sensitivity (thus acquiring more counts). The first approach is to model the Poisson noise of the SPECT acquisition in the image reconstruction algorithms; this approach led to the emergence of maximal-likelihood expectation-maximization (MLEM) algorithms [(4)], which have largely replaced the filtered-backprojection (FBP) algorithm in SPECT imaging. A related algorithm, the ordered-subset expectation maximization (OSEM) algorithm [(5)] was later developed to accelerate the reconstruction speed. The second approach is the Bayesian reconstruction [(6)] that uses prior information in the reconstruction to control the noise. The prior information can be a simple assumption that the image is smooth locally or can be a CT image of the same patient. The third approach is applying post-filtering to the images after reconstruction [(7)]. The Astonish software package developed by Philips Medical Systems provides powerful control of the resolution recovery technique and an application-specific optimization of noise suppression. It is an OSEM (Ordered Subsets Expectation Maximization) iterative method that incorporates statistical noise reduction [(8)].

Tomographic contrast is an important indicator of how well a system is performing with respect to detection of small lesions. Here it is defined as follows; Place a sphere of known size within a volume containing a uniform concentration of activity. After reconstruction, estimate the value for background count of pixels for the reconstructed image in the neighborhood of the sphere, but outside the region corresponding to the sphere. Estimate also the value of pixels within the region corresponding to the sphere for sphere count. Many other possible definitions exist and have been employed. However, the fundamental concept is to estimate the ability of the system to detect a known change in activity concentration, for a given size of a spherical object. In particular, contrast is very dependent on the size of the lesion used to estimate it. Tomographic contrast is important in that it determines the detectability of small lesions. It is affected by many different properties of the system, in particular energy resolution, the contribution of scatter and the reconstruction filter. Tomographic contrast decreases as the size of the object becomes comparable to or smaller than, the spatial resolution of the system, or when the object only partially fills the reconstruction slice. These two effects are called the partial slice filling effect and partial volume effect respectively. [(2)]

As a multicenter testing on the clinical application of Astonish iteration algorithm method. Venero et.al [(9)] studied 221 patients (94 with catheterization, 18 low likelihood for Coronary Artery Disease (CAD) and 109 with known CAD) from 3 nuclear cardiology laboratories having clinically indicated rest/stress Tc-99m sestamibi or tetrofosmin SPECT. Acquisition followed ASNC guidelines (64 projections, 20-25 seconds). Processing of the full data set included filtered back projection (FBP) and Astonish Full time Acquisition (FTA). 32 projection data sets were created by full data set stripping and processed with Astonish Half Time Acquisition (HTA). A consensus interpretation of 3 blinded readers was performed for image quality, interpretative certainty and diagnostic accuracy. They found that stress and rest perfusion image qualities (excellent/good) were 87.8%/83.3% (FBP), 97.7%/95.9% (FTA) and 96.8%/95.5% (HTA) respectively ($p <0.001$). Interpretative certainty and diagnostic accuracy were similar with FBP, FTA and HTA. Thus Full-time and half-time acquisition Astonish with simultaneously acquired line source AC improved image quality and interpretative certainty while preserved sensitivity, specificity, and normalcy rates. Therefore, half-time acquisition with AC may enhance laboratory efficiency without sacrificing image quality or diagnostic accuracy

OBJECTIVES

The aim of the study is to compare the contrast value of the images resulting from filtering with different available iteration methods with the new depth dependant collimator resolution iteration technique (Astonish) in SPECT imaging.

MATERIALS AND METHODS

A comparison was made of the calculated contrast values resulting from reconstruction for each set of iterations from different iteration methods using T-test statistical method .

A dose of 20 mCi of Tc-99m filled in Jaszczak SPECT phantom then filled completely with water and then the filling pores were closed firmly so that no activity will release from it. The study was done using (Forte) SPECT dual head gamma camera (Philips) with JET STREAM processing software package.

The acquisition parameters were prepared as follows: Extrinsic dual head using LEHR parallel-hole collimator, 60 frames (30 per each detector), Circular orbit programmed motion, step and shoot type of motion, Full field of view, 20 seconds per frame, and the matrix size is set to be 64X64.

The data was collected from the acquisition computer to the processing unit to be handled. The projection results from the acquisition were reconstructed to get the transverse slices represent each part of the phantom. The transverse slice that represents the spheres with different diameter is chosen, and then ROI is drawn on the 2 cm diameter sphere to measure the contrast against the same size ROI in the Background area within the same section.

Different iteration methods were applied for reconstruction beginning with MELM with iteration starting from 1-15 iterations, then with 3D OSEM using also iterations from 1-15 iterations and at last using Astonish with iterations from 1-15 iterations. The post filter was fixed for all methods to be Butterworth filter cutoff value = 0.35 and of order =4.0. and the subsets for all iterations are also fixed and equal=2.

Using the program provided with the processing computer for calculation of the average counts in a certain area containing certain number of pixels the transverse slice resulted from the raw data reconstruction was chosen to show hot radioactive spots within the cold (Perspex) background. The average counts for the hot spot pixels and the average counts for the same number of pixels for the background area were calculated. The contrast values were calculated for each iteration by the following formula:

$$C = \frac{\text{(Ave. sphere counts – Ave. Bkg counts)}}{\text{(Ave. sphere counts + Ave. Bkg counts)}}$$

RESULTS AND DISCUSSION

Figure (1): shows the sphere count in relation to the background count using the Astonish iteration method with different number of iteration, which shows the decrease in the background count with increasing the number of iteration.

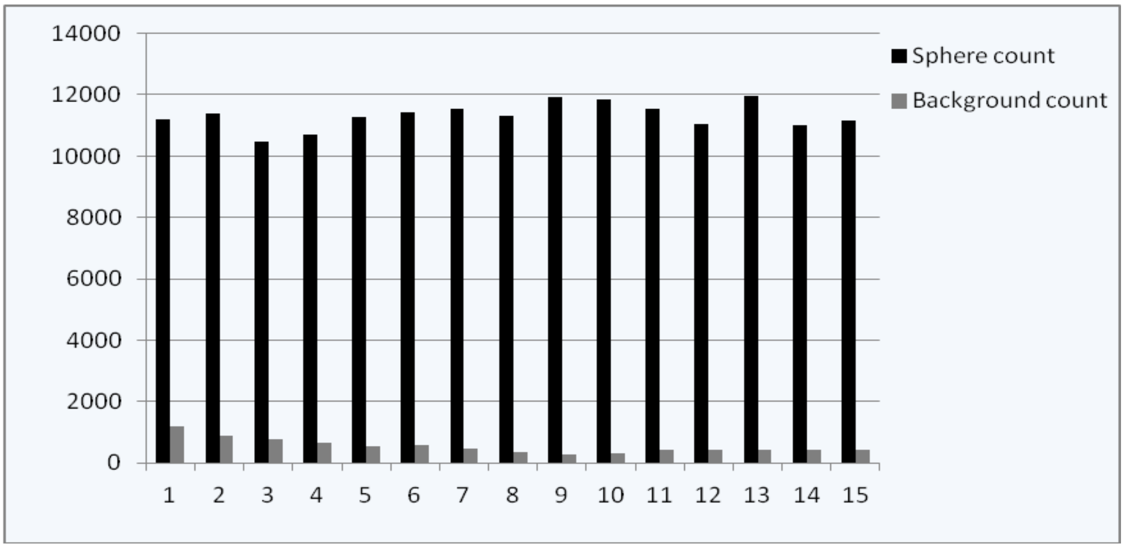

Figure 1: The number of iteration against the sphere count and background counts using Astonish method.

Figure (2): shows the sphere count in relation to background count with different number of iteration using the 3D OSEM iteration method, and it shows the stability of the background counts independent on increasing the number of iteration.

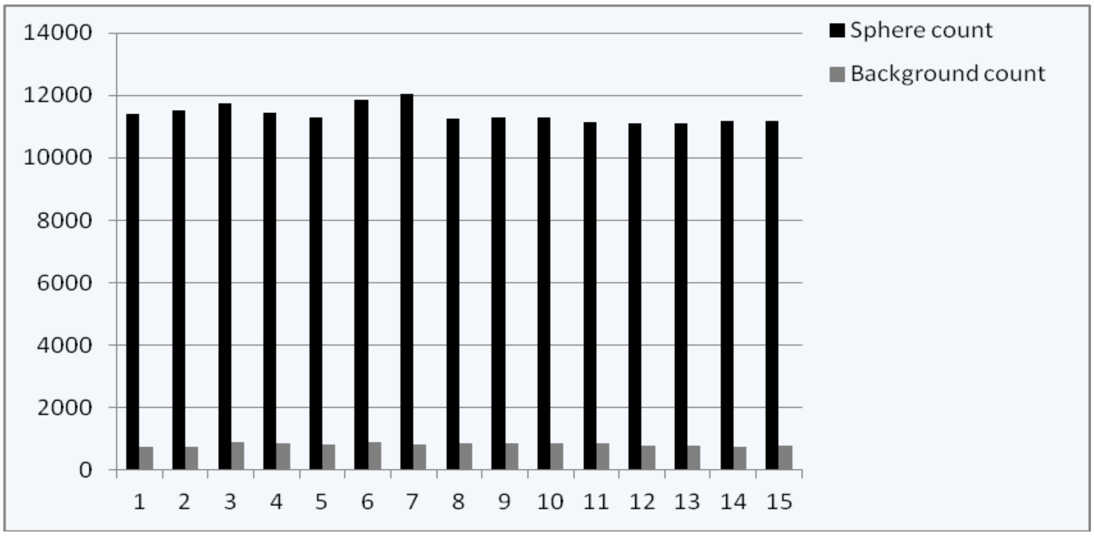

Figure 2: The number of iteration against the sphere count and background counts using 3D OSEM method

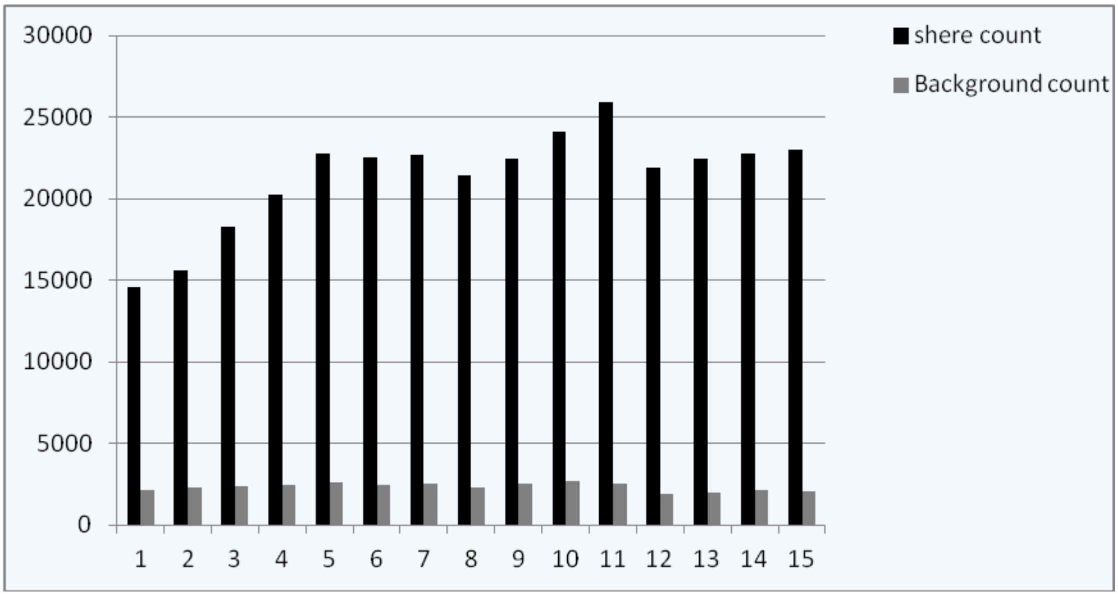

Figure 3: The number of iteration against the sphere count and background counts using MLEM method

Figure (3): shows the sphere count in relation to background count with different number of iteration using the MLEM iteration method, and it shows a visible increase of the sphere count with increasing the number of iteration, there is also an increase of the background counts independent of increasing the number of iteration.

The following charts illustrates the contrast results for each iteration starting from 1 to 15 iterations for the Astonish, 3D OSEM and MLEM reconstruction methods shown in figure (4).

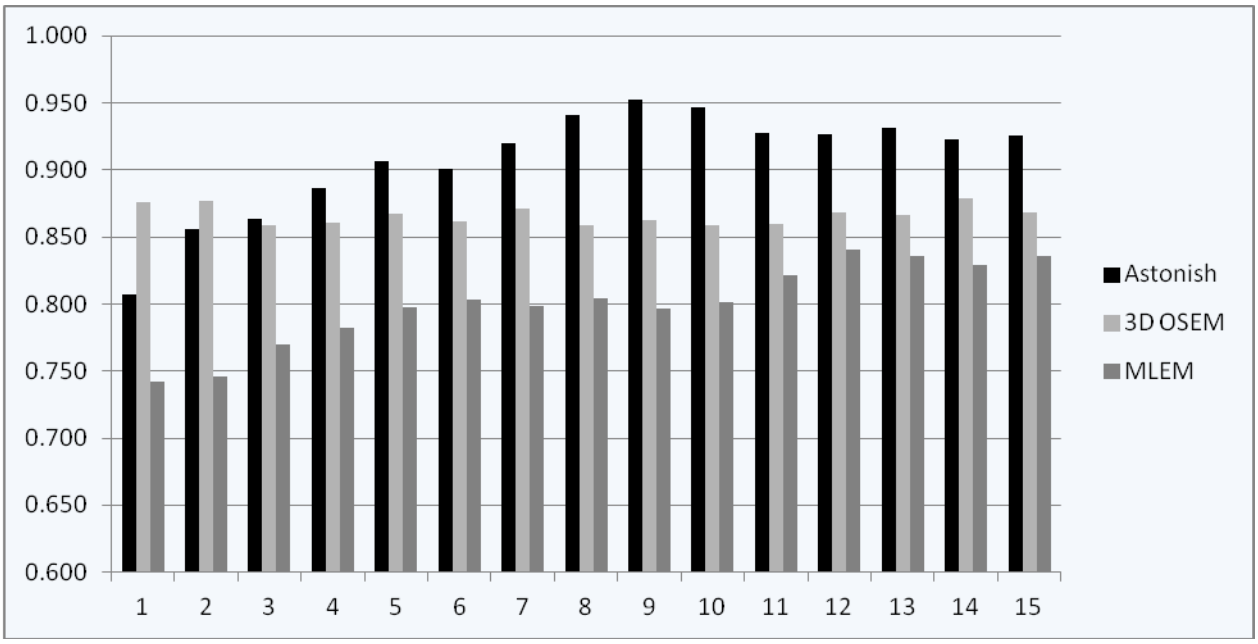

Figure 4: Comparison of the contrast results for each iteration for each reconstruction method.

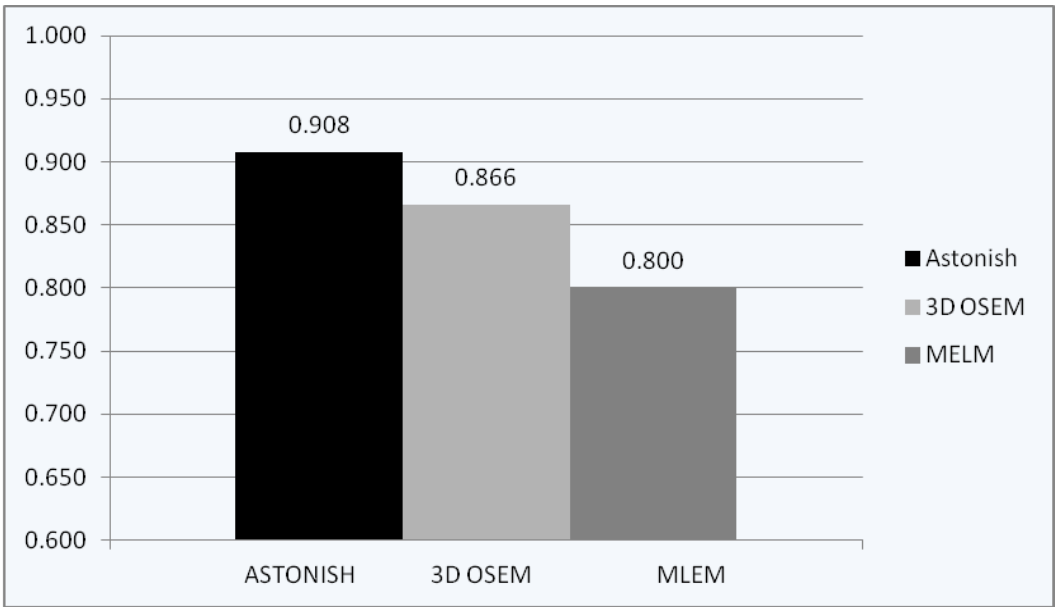

Figure 5: comparison for the average contrast result for different reconstruction method

The average contrast value for each iteration method is represented in **Figure (5)**.

Regarding the MLEM iteration method the average contrast value was [0.8002 ± 0.0304] with maximum value of 0.84 and minimum of 0.74. For 3D OSEM iteration method the average contrast value was [0.8662 ± 0.0068] with maximum value of 0.87 and minimum of 0.859. As regards to the Astonish iteration method the average contrast value was [0.907 ± 0.0394] with maximum value of 0.952 and minimum of 0.807.

There was a statistically significant improvement for the contrast value between the MLEM and 3D-OSEM iteration methods (p = 0.000001), and also there was a statistically significant improvement in the contrast value between 3D OSEM and Astonish iteration methods (P= 0.00109). Regarding the statistical relation between the Astonish and MLEM iteration methods there was a statistically significant improvement in the contrast value (P= 0.0000001)

A comparison was made between the Astonish, 3D OSEM and MLEM iteration methods on their impact in improving the contrast value of the studied phantom , it was clear that there is a statistical significant difference in contrast value between the studied iteration methods (p= 0.0000001).

The filter used for clinical applications is highly dependent upon the applications themselves, and needs to be optimized based on the specific applications [(8)]**.** Although MLEM (or OSEM) reduces image noise significantly compared with FBP or other analytical reconstruction techniques, it does not reduce the noise in SPECT images to a level that meets clinical expectations. Other approaches are usually incorporated to further suppress the noise. Therefore, one of the most commonly used techniques nowadays in SPECT image reconstruction is MLEM (or OSEM) reconstruction followed by post filtering. For SPECT image resolution recovery, a key consideration is the modeling of the resolution degradation factors, i.e., the depth-dependent resolution of the system. One approach that deblurs the acquired data models the resolution at an averaged distance. It under-recovers the resolution for objects beyond that average distance, and over-recovers the resolution for objects within the average range. Another approach that uses the frequency-distance principle (FDP)[(6)] can model the depth-dependent resolution of the system. However, the integration of this approach is not easy with two other major aspects of SPECT imaging, i.e., non-uniform attenuation and model-based scatter corrections. Still another approach is to use the convolution method in iterative reconstruction[(10)] to model the varying resolution at different distances from the detector. A similar but more efficient approach uses a slice-by-slice blurring model that was developed by Zeng *et.al.*[11]. this approach allows for depth-dependent resolution recovery as well as easy integration with non-uniform attenuation and scatter corrections. In theory, one can achieve perfect resolution recovery by correctly modeling the depth-dependent resolution of SPECT systems. However, resolution recovery alone does not make a clinically useful image. To make the image clinically useful, sufficient noise suppression is also critical. Since noise suppression approaches degrade the image resolution, a good balance of resolution recovery and noise suppression is desirable. And depth-dependent resolution recovery through internal modeling of the imaging physics. When an attenuation map is present, Astonish can additionally perform Compton scatter correction and attenuation correction.

CONCLUSION

The MLEM iteration method increases the sphere count with increasing the number of iteration; and also it increases the background counts about 3 fold than that of the 3D OSEM method, and about 4 folds that of Astonish method. While both 3D-OSEM and Astonish methods gives sphere counts that are almost within the same range but incorporating the depth dependant resolution recovery parameter in the Astonish method leads to decrease in the background count with increasing the number of iteration, Which in turn enhances the contrast of the resultant image from Astonish iteration method rather than that of both MLEM and 3D-OSEM methods. Further quantitative evaluation for this iteration technique is needed on clinical application not only for cardiac scans but also for brain scans.

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