

Influence of Iterative Reconstruction Algorithms on PET Image Resolution

G. E. Karpetas¹, C. M. Michail², G. P. Fountos², I. G. Valais², D. Nikolopoulos³, I. S. Kandarakis² and G. S. Panayiotakis¹

¹ Department of Medical Physics, Faculty of Medicine, University of Patras, 265 00 Patras, Greece

² Radiation Physics, Materials Technology and Biomedical Imaging Laboratory, Department of Biomedical Engineering, Technological Educational Institute of Athens, Egaleo, 122 10 Athens, Greece

³ Department of Computer Electronic Engineering, Technological Education Institute of Piraeus, Greece, Petrou Ralli & Thivon 250, 122 44, Aigaleo, Athens, Greece

e-mail: kandarakis@teiath.gr

Abstract. The aim of the present study was to assess image quality of PET scanners through a thin layer chromatography (TLC) plane source. The source was simulated using a previously validated Monte Carlo model. The model was developed by using the GATE MC package and reconstructed images obtained with the STIR software for tomographic image reconstruction. The simulated PET scanner was the GE DiscoveryST. A plane source consisted of a TLC plate, was simulated by a layer of silica gel on aluminum (Al) foil substrates, immersed in 18F-FDG bath solution (1MBq). Image quality was assessed in terms of the modulation transfer function (MTF). MTF curves were estimated from transverse reconstructed images of the plane source. Images were reconstructed by the maximum likelihood estimation (MLE)-OSMAPOSL, the ordered subsets separable paraboloidal surrogate (OSSPS), the median root prior (MRP) and OSMAPOSL with quadratic prior, algorithms. OSMAPOSL reconstruction was assessed by using fixed subsets and various iterations, as well as by using various beta (hyper) parameter values. MTF values were found to increase with increasing iterations. MTF also improves by using lower beta values. The simulated PET evaluation method, based on the TLC plane source, can be useful in the resolution assessment of PET scanners.

Keywords: PET; MTF; image quality; iterative reconstruction; Monte Carlo

1. Introduction

The most common image quality metrics in order to assess spatial resolution in Positron Emission Tomography (PET) scanners are the Point Spread Function (PSF) and the corresponding Full Width at Half Maximum (FWHM) obtained from point sources [1]. Only in a few studies spatial resolution was assessed in terms of the Modulation Transfer Function (MTF), calculated from PSF data [2]. The same method was employed in studies concerning combined micro PET/CT scanners, in which MTF was estimated in the CT detector of the combined system [3]. MTF can be also assessed from a line source through the estimation of the Line Spread Function (LSF). The use of the LSF method for determining the MTF in tomographic imagers was initially introduced by Boone [4] who applied this method for CT scanners evaluation. Fountos *et al.* [5,6] recently introduced a similar method for SPECT scanners by immersing a medical X-ray film in a solution of dithiothreitol (DTT)/Tc-99m(III)-DMSA to obtain the MTF. The purpose of the present study was to extend a previous validated Monte Carlo model [7-9] in order to obtain PET image resolution through the estimation of the MTF. Within the context of the model, MTF was estimated by the simulation of a thin layer chromatography (TLC) plane source filled with 18F-FDG, using a software model based on the GATE Monte Carlo package. The GATE package,

which was developed by the Open-GATE collaboration, is an open-source extension of the Geant4 Monte Carlo toolkit [10]. GATE was used in combination with the STIR image reconstruction software [11] to obtain the plane source reconstructed images. The influence of different number of iterations and subsets, as well as the influence of various beta (hyper) parameter values, in the iterative image reconstruction was also investigated. The simulation of this plane source phantom provides an accurate model that is useful to fully characterize the performance of nuclear medicine imaging systems.

2. Materials And Methods

1.1 Geometry of the modeled PET scanner

The Discovery ST PET/CT system was modeled. The system incorporates Bismuth Germinate Oxide (BGO) [12,13] crystals with dimensions of 6.3x6.3x30 mm. The detector ring is finally comprised of 35 modules, i.e. 280 crystal blocks (for a total of 10080 BGO crystals). The dimensions of the rings are 88.6 cm diameter with a 15.7 cm axial and 70cm transaxial fields of view (FOV).

2.2. Software for simulation

The employed simulation software was the GATE toolkit. In every BGO crystal of the detector block, a Gaussian energy blur is applied with an average energy resolution of 17%, referenced at 511 keV. Afterwards, a 300 ns dead time value was applied on the single events in the BGO crystal [14] by a paralyzable Deadtime module. The Quantum Detection Efficiency (QDE) [15,16] of the crystals was found 0.94 at 511 keV. The sinogram output file (.ima) was used by STIR as input file for the reconstruction of the simulated plane source image. The coincidence time window was set to 11.7 ns.

2.3. Preparation of the MTF test object

The MTF test object, which was simulated in this study, was a planar source similar to those used in previous works of Boone [4] for CT, Fountos *et al.* [5,6] for SPECT systems and Karpetas *et al.* for PET systems [9]. The plate was simulated with a layer of silica gel on Al foil substrates. The dimensions of the TLC plate were 5x10 cm² and it was assumed to be immersed in an 18F-FDG bath solution (1 MBq). The MTF test object was simulated between two semi-cylindrical polyethylene blocks with 20 cm diameter and 70 cm length. The importance of the hyper parameter (beta value) (which controls the weight of the penalty term and the importance of the prior) was also examined in order to investigate the corresponding influence on the MTF [17].

2.4 Modulation Transfer Function (MTF)

MTF was obtained, by using the LSF method. The thin plane source was simulated at a slight angle of 3° with respect to the horizontal or vertical axis, following the technique proposed by Fujita to avoid aliasing effects [18]. The final LSF was obtained by averaging all line LSF profiles after angle correction. The angle correction was performed following the procedure described in Fountos *et al.* [5].

3. Results and Discussion

3.2 Modulation Transfer Function (MTF)

Figure 1 shows the influence of the variation of the hyper parameter on the Maximum Likelihood Estimation (MLE)-OSMAPOS, the Ordered Subsets Separable Paraboloidal Surrogate (OSSPS), which is a relaxed preconditioned sub-gradient descent algorithm, the Median Root Prior (MRP) and OSMAPOS with quadratic prior, algorithms. The Median Root Prior is obtained by using Median as data processor. This prior is an extension of the idea first developed for the Median Root Prior which takes any Data Processor (i.e. a filter), and computes the prior gradient [19]. The hyper parameter showed greater influence in the MTF values when the OSMAPOS with quadratic prior when 20

iterations was used, while the influence of beta prior was diminished with the OSSPS with 3 iterations. Figure 2 shows the resulted MTF curves when the hyper parameter is allowed to vary from a value close to zero up to 3 with various iterations and subsets, in order to investigate its impact on PET image resolution through the MTF. As it can be seen, spatial resolution is improved by using lower beta values [17]. Larger beta values result in smoother images, while lower values decrease the importance of the penalty term [17, 20]. By increasing the number of iterations, the influence of the beta parameter on image resolution tends to decrease showing smaller deviations. For example, in the 15 subsets/2 iterations image (30 MLEM equivalent iterations), deviations of the order of 60.2% are shown between the 0.01 and 3 beta value, whereas in the 300 MLEM equivalent iterations (15 subsets/ 20 iterations) image the corresponding difference decrease to 31.3%, at the spatial frequency of 0.05 cycles/mm. Image quality in PET scanners is affected by a large number of factors including the number of iterations and subsets in ordered subset-type algorithms, filter in filtered backprojection, and post-filtering [21, 22]. In filtered backprojection reconstructed images, resolution can be assessed in a straight manner, which is not the case for OSEM images, available in most clinical systems, since iterative reconstruction algorithms are non-linear and difficult to evaluate [23]. By increasing the number of iterations, within a reasonably wide range of practically encountered clinical situations [24], resolution improves and tumors appear brighter. However, a point is reached after which resolution does not improve [21]. The hyper parameter which controls the weight of the penalty term (beta value) and the importance of the prior, has an influence also on both image resolution and noise. Spatial resolution is improved by using a lower beta value at the expense of higher image noise. In the work of Cheng *et al* [17] resolution stabilized after the value of 20 iterations. Larger values of b yield smoother images, while lower values decreases the importance of the penalty term and yield a solution closer to the MLEM algorithm [17, 20].

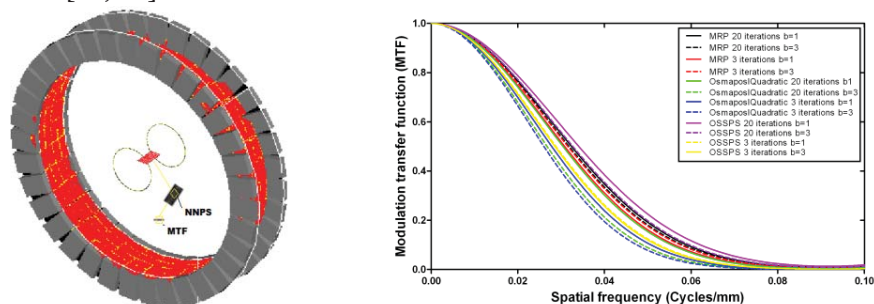


Figure 1. Simulation setup (left). Comparison between the MTFs obtained from various iterative algorithms using various values of the beta (hyper) parameter (right).

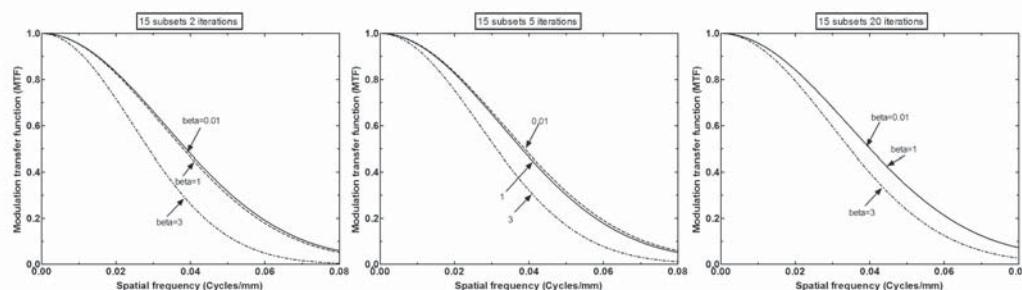


Figure 2. Comparison between the MTFs obtained from the OSMAPSL with 30, 75 and 300 MLEM iterations (15 subsets, 2, 5 and 20 iterations) reconstructed images using various values of the hyper parameter.

4. Conclusions

Image quality characterisation of a PET scanner was achieved through the estimation of the MTF of thin ¹⁸F-FDG plane source images, simulated by a Monte Carlo model. The influence of iterative image reconstruction on the plane source simulated images was investigated by using fixed subsets and various iterations. Image quality was found to improve by increasing the number of iterations (number of image updates). Spatial resolution is also improved by using lower beta values. Larger beta values have as a result smoother images, while lower values decreases the importance of the penalty term. The method was used for the image quality assessment and optimisation, but it can be also useful in research for the further development of PET and SPECT scanners through GATE simulations.

5. References

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