Optimizing Positron Emission Tomography radiation dose using Monte Carlo N-Particle simulations

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Abstract

Positron emission tomography (PET) is a functional imaging procedure that uses a radioactive tracer to visualize organ or tissues functions for the purposes of diagnoses, staging and therapy monitoring of cancer. However, reconstructed images produced by PET are often accompanied with noise resulting in inaccuracies in diagnoses and other treatment procedures. This has been more apparent with patients who are grossly overweight or morbidly obese. The present study was conducted to quantitatively determine the optimal amount of injected radiotracer required to attain sufficient quality of PET images based on differing physiques. A simulation process was undertaken and completed in three main phases: (i) modelling of four different sizes of phantoms typifying underweight, normal, overweight and obese patients using Monte Carlo N-particle (MCNP); (ii) reconstructing images in the range of 435-650 keV energy window using MATLAB programming platform; and (iii) determining optimal dose of activity required to induce adequate tumor signal-to-noise ratio (SNR) value for a proposed imaging system. Three different activities were tested with modelled National Electrical Manufacturers Association (NEMA) International Electrotechnical Commission (IEC) PET phantom starting from 6, 8 and 10 mCi of ¹⁸F-fluorodeoxyglucose (FDG) with tumor-to-background ratio (TBR), ranging from 2:1 to 6:1. Results showed that reducing activity of injected radiotracer in PET imaging was possible for variations in body weights while maintaining image quality. The SNR value for different dosages differed only by maximum value less than 2.0. The present study serves as a basis for further clinical PET studies using reduced radiotracer doses as compared with conventional PET protocols.

Keywords: Body Mass Index; Monte Carlo N-particle; Positron Emission Tomography; Tumor-to-Background Ratio
1.- INTRODUCTION

In early detection, diagnoses and treatments of major health conditions such as cancer, the highest quality of medical images of the affected interior of the body is most crucial for the primary purpose of clinical analyses or medical interventions by an attending physician. When an accurate diagnosis is expected, medical images need to be extremely accurate in portraying the anatomical and metabolic conditions of the affected areas. To this effect, quality images provided by PET in nuclear functional imaging technology has been greatly coveted as an aid in the diagnosis of cancer. However, PET images often have to be corrected or rectified for a number of degrading factors such as attenuation, scatter, and randoms [Townsend 2004].

Photons attenuation, the progressive loss of energy, generally occurred when annihilation events were removed from the line of response (LOR). Meanwhile, scatter photons are photons that scatter in tissues prior to coincide with photon detection. These effects contribute much to background noise of PET images which are known to have linear dependent on patients’ body size [Karakatsanis and Nikita 2009; Tatsumi et al., 2002; Turkington et al., 2004]. The probability of a photon being attenuated or impaired is known to increase as the depth of interaction increases, while the likelihood of photon escape without scattering increases with decreases in depth of interaction. Body mass index (BMI) is well known to reveal significant positive correlations with image noise due to increases in photon attenuation and scatter fraction [Halpern et al., 2005; Lindemann et al., 2018; Taniguchi et al., 2015; Turkington et al., 2004; Yan et al., 2016].

The degradation of quality in PET imaging has been cited to be closely related to low-energy photons in reconstructed images as well as fewer number of counts registered in the counting system [Musarudin et al., 2015; Wickham et al., 2017]. A number of procedures have been proposed to resolve these problems, which included optimizing acquisition time [Carlier et al., 2014; Lindemann et al., 2018; Wickham et al., 2017], optimizing energy threshold [Musarudin et al., 2015], introducing time-of-flight (TOF) in PET imaging
One approach in tumor detection by means of manipulating the amount of injected dosage in PET imaging was proposed with the primary objective of obtaining high image quality. It was reported that bigger size patients, such as overweight and obese patients, caused more coincidence photons losses to occur due to high scattering events and photon attenuation [Fakhri et al., 2007; Groot et al., 2013; Saade et al., 2019]. In studies by [Ferrero et al., 2011; König et al., 2012], high percentage of scatter fraction was reported for various sizes of human phantom diameter. Correspondingly, a linear relationship was reported between BMI and total attenuation of patients [Fakhri et al., 2007]. A quadratic relationship was also reported between human phantom’s diameter and scatter fraction [König et al., 2012]. Prieto et al., [2018] cited that a common practice to confront the effects of a patient’s body in PET imaging was to initiate higher FDG dose per body weight in order to maintain image quality. The aim of this procedure was to detect metabolically active malignant tumors which are cancerous. Linear relationships between patients’ body mass and prescribed doses have been documented by the European Association of Nuclear Medicine (EANM). However, clinical practices have confirmed that the linear relationships were incapable of yielding good quality images, particularly in obese patients.

Against these backgrounds, the present study was conducted with one primary objective: to determine the optimal amount of injected radiotracer for PET on the basis of different phantom simulation dimensions while maintaining sufficiently high image quality.

2.- MATERIALS AND METHODS

In the present study, the procedure used in developing a realistic model of a PET system included a combination of MCNP and MATLAB programming platform to model Siemens Biograph TruePoint PET/CT scanner. The procedures started with the design of PET
camera and phantom geometry for determining patients’ categories, followed by image reconstruction and evaluation.

2.1.-PET camera and phantom modelling

Conventionally, PET is a circular tomograph with separate detectors. Based on Siemens Biograph TruePoint PET/CT, 48 block detectors were arranged in a rotational position, consisting of a Lutetium oxyorthosilicate (LSO) crystals and an array of photomultiplier tubes (PMTs) [Jakoby et al., 2009]. Since the detectors were arranged in a rotational position, the geometry for each detector needed to be designed in different angles. The feature of repeated structure card in MCNP made this possible by defining the surfaces and cells followed with transformation card [X-5 Monte Carlo Team, 2003]. Figure 1 shows the simulated PET scanner with detectors in different angles and its dimension is shown as in Table 1. In the present study, only single-ring detector was developed while the real Biograph PET/CT had 39 detector rings. This limitation was for the purpose of reducing the complexity of the simulation and yet still preserving the accuracy of its performance.

Figure 1.- Biograph TruePoint PET/CT using MCNP in (a) xz direction (b) xy direction.
Table 1.- Specification Biograph of TruePoint PET/CT.

<table>
<thead>
<tr>
<th>PET camera</th>
<th>Biograph TruePoint PET/CT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Crystal dimension (mm$^3$)</td>
<td>4.0 × 4.0 × 20</td>
</tr>
<tr>
<td>Detector ring diameter (mm)</td>
<td>842</td>
</tr>
<tr>
<td>Detector per ring (sub-detectors)</td>
<td>624</td>
</tr>
</tbody>
</table>

Following the above-mentioned procedure, the geometry of phantom with four different sizes was defined representing underweight, normal, overweight and obese patients. The geometry of the phantom was based on NEMA IEC PET phantom (Figure 2) which involved some calculations in order to obtain a realistic and comparable model as the real phantom.

Assuming the height of a patient was 175 cm, the weight of these phantoms can be represented by Equation 1 [Groot et al., 2013]:

$$w = v \times \rho$$  \hspace{1cm} (1)

where $w$, $v$ and $\rho$ are the weight, volume and density.

Figure 2.- Dimension of NEMA IEC PET phantom.
In addition, BMI of the modelled phantom could be calculated using Equation 2:

\[ \text{BMI} = \frac{w}{h^2} \]  

(2)

Based on Equations 1 and 2, the four phantoms modelled correspond approximately to body mass and BMI as shown in Table 2.

<table>
<thead>
<tr>
<th>Diameter</th>
<th>Weight (kg)</th>
<th>BMI</th>
<th>BMI category</th>
</tr>
</thead>
<tbody>
<tr>
<td>Major</td>
<td>Minor</td>
<td></td>
<td></td>
</tr>
<tr>
<td>21</td>
<td>16</td>
<td>48.25</td>
<td>15.76</td>
</tr>
<tr>
<td>25.2</td>
<td>19.2</td>
<td>69.48</td>
<td>22.69</td>
</tr>
<tr>
<td>27.3</td>
<td>20.8</td>
<td>81.57</td>
<td>26.6</td>
</tr>
<tr>
<td>29.4</td>
<td>22.4</td>
<td>86.59</td>
<td>28.27</td>
</tr>
</tbody>
</table>

Subsequent to designing of PET camera and phantom, the total number of photons emitted by the source was calculated. The total number of photons was based on several parameters including phantom volume, activity density, TBR and time of imaging as in Equation 3:

\[ \text{NPS} = \text{activity} \times \text{volume} \times \text{time} \]  

(3)

Based on Equation 3, each of phantom volume with different amount of dosage yielded the number of photons as in Table 3.

<table>
<thead>
<tr>
<th>Dosage (mCi)</th>
<th>Number of photons</th>
</tr>
</thead>
<tbody>
<tr>
<td>6</td>
<td>(2.664 \times 10^{10})</td>
</tr>
<tr>
<td>8</td>
<td>(3.552 \times 10^{10})</td>
</tr>
<tr>
<td>10</td>
<td>(4.440 \times 10^{10})</td>
</tr>
</tbody>
</table>
2.2.- Image reconstruction
Following PET camera and phantom modelling procedures, PET data were reconstructed, firstly, by tracking the coincidence photons from a list-mode output data called Particle Track Output Card (PTRAC) generated by the PET MCNP code. Secondly, the amount of energy deposited in the detector material was calculated which generally contributed from absorption, scattering and termination events. Upon completion of energy calculation, an Anger Logic Algorithm was used to determine the actual point of interaction which was the number of crystal elements [Guérin and Fakhri 2008], LOR to be defined. In real PET imaging, several physical effects were involved in the process of photon detection such as statistical fluctuations which led to energy broadening. To simulate this condition, the effect was modelled using a full-width half maximum (FWHM) Gaussian distribution, as demonstrated in studies by [Guérin and Fakhri 2008]. In the present study, the default energy threshold level was set at 435 - 650 keV.

To all intents and purposes, PET is used fundamentally to obtain many projections from different angles around an object being imaged. In image reconstruction, a number of projections were generally acquired at equally spaced angles between 0 to 360 degrees. Sinograms were subsequently obtained from each row of the projection image using a filter, such as Hann, in the process. To reconstruct the slices of object, a simple back projection algorithm was used with matrix size of 168 x 168. At this stage, the acquired image, also known as a slice, was reconstructed from raw projection data using filtered-back-projection (FBP) method.

2.3.- Performance evaluation
Subsequent to procedures previously described, PET image analysis was performed based on SNR measured on the reconstructed image. This is to quantify the performance of PET scanner in terms of visibility. Literature has it that there were several methods of calculating SNR on PET image analysis. In the present study, the approach proposed by [Lois et al., 2010] was adopted. Tumor SNR was calculated by finding the
between tumor signal and background and comparing it to noise in the background as in Equation 4:

\[
SNR = \frac{\text{Signal} - \text{Background}}{\text{Noise}}
\]  

(4)

Both tumor signal and background were in terms of mean value in the region of interest (ROI). Meanwhile, noise was defined by the standard deviation of the pixel values calculated in the background ROIs. Masuda et al., [2009] cited that noise estimation based on ROI method could be affected by location and size of ROI, as well as the uniformity of the image. The present study defined ROI of tumor and background as presented in Figure 3.

![Image of ROI sections](image)

Figure 3.- The ROI defined for SNR computation. (a) Tumor ROI (b) Background ROIs

3.- RESULTS

At the conclusion of the present study, the relationships among patients’ categories, optimum dosage and TBR were established. The SNR values for four phantoms at five different TBRs are shown in Figures 4 a, b, c, d and e. The optimal dosage for each BMI category and TBR were obtained from the values of maximum SNR.
(a) SNR for TBR 2:1

(b) SNR for TBR 3:1

(c) SNR for TBR 4:1
Figure 4.- SNR measured for various phantom and TBRs, (a) SNR for TBR 2:1, (b) SNR for TBR 3:1, (c) SNR for TBR 4:1, (d) SNR for TBR 5:1, (e) SNR for TBR 6:1.

4.- DISCUSSION

Degradation in the values of SNR was observed as phantom diameter increased. Such effects were observed for all TBRs modeled. Results in the present study suggest that the contribution of noise was more prominent when imaging a phantom of bigger size. Noise resulting from scatter events was observed to increase with increments in phantom size. The observation suggested that phantom mass affected the overall performance such that more scatter events were recorded with more true events being lost. This could be taken as an early indicator of annihilation position detection displacement.
Small improvements in SNR were consistently observed when increase in dosages were performed. This could be due to more counts being recorded for PET image reconstruction. However, implementation of dosage above optimal value caused significant decrease in SNR values as clearly seen for overweight category at all TBRs. This could be due to more counts being involved which primarily came from scatter events when implementing relatively high dosages. The trend was also observed at lower TBRs indicating that at early stage of tumor, implementation of optimal dosage was crucial in improving tumor visibility.

**5.- CONCLUSIONS**

In the present study, the optimal dosage based on patients’ size was determined by evaluating image quality using SNR analysis. This was executed from early stage of tumor (TBR 2:1) known to be the most difficult stage to diagnose its existence in a patient’s body. Based on SNR analyses for four different categories of patients, the study concluded that increase in dosage did not automatically improve the quality of PET images. Other factors such as the amount of scatter events with true counts lost at different phantom mass need to be considered in order to get the optimal signals for the respective phantom imaging.

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