RECONSTRUCTION OF THE NEMA IEC BODY PHANTOM FROM J-PET TOTAL-BODY SCANNER SIMULATION USING STIR*

P. Kopka, K. Klimaszewski

for the J-PET Collaboration

National Centre for Nuclear Research, 05-400 Otwock-Świerk, Poland

(Received November 18, 2019)

The total-body positron emission tomography (PET) with its large field of view (FOV) brings many benefits *i.a.* scan time reduction, increase of the sensitivity and simultaneous imagining for multiple organs. However, introduction of oblique lines of response (LORs) can result in image quality degradation. In this paper, we provide a study of such effects based on simulations of two-meter long J-PET scanner. Simulation of a point source shows a degradation of the axial resolution due to parallax effect by factor of four. However, based on a simulation of the NEMA IEC-BODY phantom, we do not observe a significant reduction of image quality.

DOI:10.5506/APhysPolB.51.357

1. Introduction

Current clinical PET scanners have an axial FOV of 15–25 cm in length, which leads to long total imaging time and limited body coverage. The purpose of the Jagiellonian PET (J-PET) project is to find a cost-effective way of the total-body PET imaging using polymer detectors and to elaborate an *in vivo* positronium imaging of tissue pathology [1, 2]. The cost of the available, clinical PET scanners, based on crystal scintillators, increases almost linearly proportional to the FOV, whereas the cost of the J-PET scanner does not increase significantly when increasing the FOV [3, 4]. For long cylindrical PET detectors, the axial resolution can be degraded by the parallax effect between detected pairs with a large axial difference [5]. In this paper, we present the effects of increasing axial difference on scanner resolution and reconstructed image quality.

* Presented at the 3rd Jagiellonian Symposium on Fundamental and Applied Subatomic Physics, Kraków, Poland, June 23–28, 2019.

2. Material and methods

The J-PET scanner is simulated using the Geant4 application for tomographic emission (GATE) [6]. The simulated geometry forms a single barrel of 855.6 mm in diameter consisting of 384 modules [7, 8]. The modules are made of EJ230 scintillator with size of $20 \times 7 \times 2000$ mm^3. For the purpose of image reconstruction, the barrel is split into 500 axial rings 4 mm wide. The ring block width is less than the assumed detection resolution of the axial position simulated for silicon photomultipliers readout solution combined with a wavelength shifters (WLS) layer [9]. The coincidence time window was set to 3 ns and the detection energy threshold was set to 200 keV.

Results of two simulations are presented, the first simulation for a point source located in the center of the scanner shifted by 1 mm in the transaxial plane. In the second case, a NEMA IEC-BODY phantom is simulated. This water phantom consists of a body phantom, a lung insert and an insert with six spheres with various sizes. The activity for four smaller spheres is 4 times larger than the activity of the background, the middle cylinder and the two largest spheres are filled with air [10].

Image reconstruction is performed using the Filtered-Back-Projection 3D Reprojection (FBP3DRP) algorithm implemented in the Software for Tomographic Image Reconstruction (STIR) [11] with voxel size of an attenuation correction is applied in the case of the NEMA IEC-BODY phantom reconstruction.

3. Results

To test how scanner resolution depends on the LOR angle with regards to the transverse plane (α), a reconstruction of point source is performed by selection of LORs only at a specific angle. The axial resolution is degraded by the parallax effect for LORs with large angle α (Figs. 1, 2). The transax-

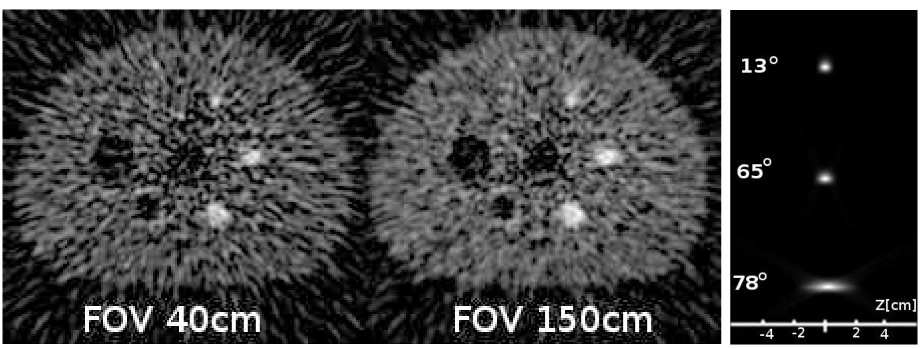


Fig. 1. Left: Reconstruction of NEMA IEC-BODY. Right: Coronal images of reconstructed point source (PDF) with different angle α.

ial resolution is independent of the angle α and approximately constant at 4.6 mm. However, the axial resolution changes in the range from 6.2 mm to 22.7 mm with a steep increase from angle α of 50°. The maximum measured $\alpha = 78°$ corresponds to the 200 cm ring difference.

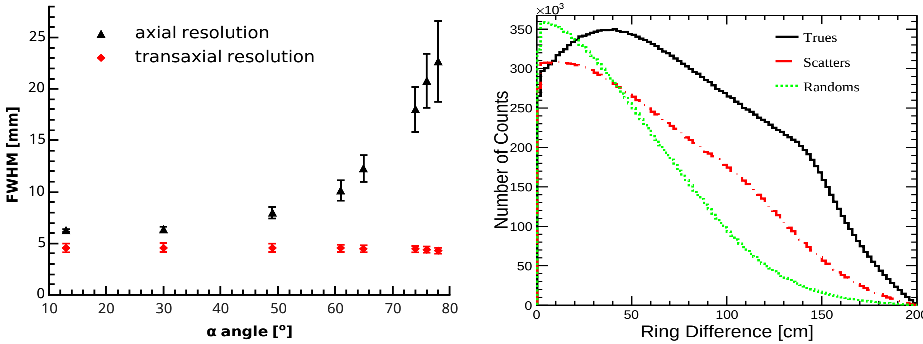


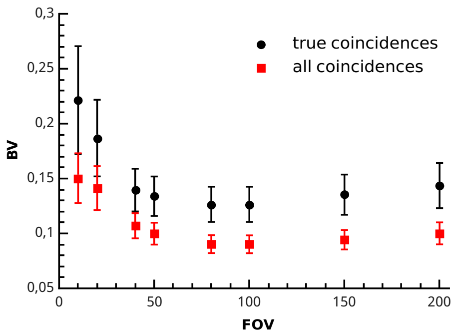
Fig. 2. Left: Resolution as a function of α angle for the point source. Right: Number of detected coincidences as a function of ring difference for NEMA ICE-BODY phantom.

To test the behaviour of image quality as a function of FOV, the simulation of NEMA ICE-BODY phantom is carried out. The images for true and all coincidences are reconstructed with FBP3DRP algorithm with varying axial FOV defined by the maximal ring difference. An example of two reconstructed images for FOV 40 cm and 150 cm from true coincidences is shown in Fig. 1.

To evaluate the image quality, we calculate the contrast recovery coefficient (CRC) and the background variability (BV) for the 22 mm hot sphere from the reconstructed images. The BV is defined by the ratio of the standard deviation to the mean in the background. The CRC is calculated for the hot sphere as

$$\mathrm{CRC} = \frac{(\mathrm{sphere\_mean})/(\mathrm{background\_mean}) - 1}{\mathrm{true\_contrast} - 1} \,.$$

The BV increases for small FOV as the sample size decreases (Fig. 3), also the CRC stabilises after 40 cm FOV. The number of detected coincidences as a function of ring difference (Fig. 2) explains the behaviour of the image quality parameters. Even though we showed that axial resolution decreases for large angles, the number of coincidences sharply decreases with FOV and no adverse effect on image quality is observed. For FOV larger than 40 cm, the image quality improves slightly, while reconstruction time for FBP3DR increases 6 fold (from 9.5 hours at FOV = 40 cm to 59 hours at FOV = 150 cm for single Intel Xeon E5-2680 v2 CPU at 2.8 GHz).



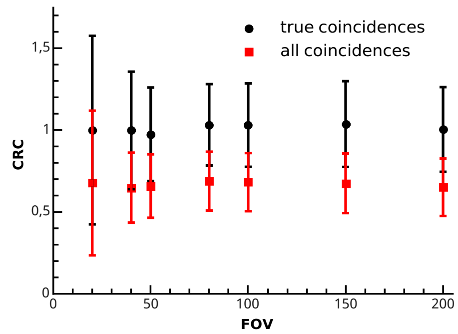


Fig. 3. Left: Background variability (22 mm hot sphere) for different FOView. Right: Contrast recovery coefficient (22 mm hot sphere) for different FOV.

4. Conclusions

The degradation of the axial resolution has to be investigated for total-body PET scanners. For the two-meter long J-PET scanner, it increases rapidly for α angles larger than 50° (102 cm FOV). At the same time, the image quality does not improve from about 100 cm FOV, while the calculation time increases quickly. Therefore, it is beneficial to consider only coincidences with α angles less than 50°. Further studies including Time-of-Flight (ToF) measurements and geometries with multiple layers will be required to optimise data selection for the J-PET scanner prototype [4].

We acknowledge support by the Foundation for Polish Science through MPD and TEAM POIR.04.04.00-00-4204/17 programmes, 380 the EU and MSHE grant No. POIG.02.03.00-161 00-013/09.

REFERENCES

[1] P. Moskal, B. Jasińska, E.Ł. Stępień, S. Bass, *Nature Rev. Phys.* **1**, 527 (2019).
[2] P. Moskal *et al.*, *Phys. Med. Biol.* **64**, 055017 (2019).
[3] P. Moskal *et al.*, *Phys. Med. Biol.* **61**, 2025 (2016).
[4] S. Niedźwiecki *et al.*, *Acta Phys. Pol. B* **48**, 1567 (2017).
[5] X. Zhang *et al.*, *Phys. Med. Biol.* **62**, 2465 (2017).
[6] S. Jan *et al.*, *Phys. Med. Biol.* **49**, 4543 (2004).
[7] P. Kowalski *et al.*, *Phys. Med. Biol.* **63**, 165008 (2018).
[8] G. Korcyl *et al.*, *IEEE Trans. Med. Imag.* **37**, 2526 (2018).
[9] J. Smyrski *et al.*, *Nucl. Instrum. Methods Phys. Res. A* **851**, 39 (2017).
[10] National Electrical Manufacturers Association, Norm NEMA-NU-2 Performance Measurements of Positron Emission Tomographs (PETs), 2012.
[11] K. Thielemans *et al.*, *Phys. Med. Biol.* **57**, 867 (2012).