FEASIBILITY STUDY OF PET IMAGING USING SINGLE-SCATTERED EVENTS WITH TOF*

Ritesh Verma, Pragya Das

Indian Institute of Technology Bombay, Mumbai — 400076, India

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A substantial amount of scattered data is rejected in a conventional Positron Emission Tomography (PET). We propose a novel algorithm for PET image reconstruction from single-scattered (inside tissue) events with a known Time of Flight (TOF) and without energy information, particularly useful for plastic scintillator. Previous single scatter reconstruction algorithms relied on both energy and time information of detected events which is applicable only for inorganic scintillators. We discuss the framework of our reconstruction algorithm with the GATE simulated data where we achieved a resolution of 1.8 cm approximately for a line source. Currently, our work is demonstrated for 2D images and the algorithm accurately predicts the source location.

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1. Introduction

An image quality of nuclear medicine scans is often degraded by various factors such as scattering within the phantom. The scatter fraction in PET typically ranges from about 15% in 2D to 40% or more in 3D [1]. Traditional PET reconstruction algorithms estimate and subtract scatter contributions from detected coincidences to enable a more accurate activity reconstruction [2]. Scatter coincidences still carry valuable information about the activity distribution. The use of the list mode format for TOF-PET systems has paved the way for us to reconstruct the activity distribution using single-scattered coincidences [3]. Previous reconstruction algorithms, as detailed by Ghosh and Das [4], relied solely on energy information, whereas Hemmati *et al.* [5], utilized both energy and time information to determine the locus of the scattering points. However, these algorithms are ineffective for plastic scintillators, where energy deposition occurs primarily through

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Compton scattering. Nonetheless, the excellent timing resolution of plastic scintillators gives the information of TOF, like J-PET [6, 7]. The proposed method was evaluated on a uniform phantom filled with water. By integrating this method with conventional reconstruction techniques of using true coincidences, the sensitivity of the system will improve. We can incorporate the scatter information into an iterative reconstruction algorithm like OSEM (Ordered Subsets Expectation Maximization) by reconstructing the image using only the true coincidence events and adding scattered coincidences into the next iterations. Since scatter events contain indirect information about the activity distribution, they can be modeled and included as a separate likelihood term in the iterative algorithm. The proposed method may provide activity distribution where true coincidence events are scarce due to tissue scattering or in low-dose PET scans. In this proof-of-concept study, we generated direct images of the source activity without utilizing conventional iterative reconstruction algorithms. We employed a point-wise imaging technique, where each event is reconstructed independently from the others. This method was selected due to its simplicity and all the studies have been conducted under precise TOF information without incorporating multiple scattering coincidences.

2. Materials and method

To determine the annihilation point from single-scattered events, we developed an algorithm based on the assumption that a photon reaches point A first and then point B after scattering at C, as illustrated in Fig. 1 (left). This method is also valid for cases where the photon reaches point B first, due to symmetry. Studies indicate that the scattering point lies on the arc of a circle in 2D (Fig. 1 (right)) and on a prolate spheroid in 3D geometry for a fixed scattering angle [8], as a result of the geometrical equations de-

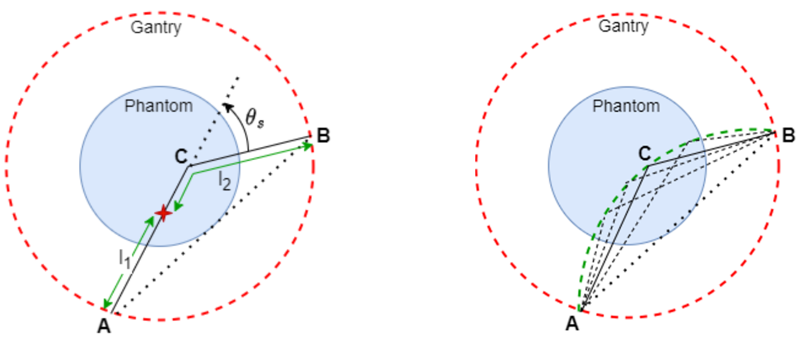

Fig. 1. Left: Photon track after a single-scattering event. Right: The scattering locus, shown in green, as a circular arc in 2D geometry. See details in the text.

rived from the inscribed angle theorem. We identified possible annihilation points for a fixed scattering angle within the phantom. Let l_1 and l_2 be the distances from the annihilation point to points A and B, respectively. The difference in the Time of Flight (TOF) provides the path difference (Δl) the photon has traveled to reach both detectors.

Here, Δl is given by the speed of light in the medium multiplied by the TOF difference. The annihilation point can then be estimated from the following set of equations:

$$\begin{aligned} l_1 + l_2 &= \mathrm{AC} + \mathrm{BC} = L\,, \\ l_1 - l_2 &= \Delta l\,. \end{aligned}$$

2.1. Circular arc modeling and reconstruction framework

We have developed a model to determine the scattering locus by constructing arcs. By creating circles with the same radius from both the detection points *i.e.* A and B, we obtained two intersection points when the radius exceeded half the distance of LOR. The circles of equal radius from these intersection points pass through points A and B. By iterating the radius, we generated multiple scattering points representing different scattering angles. The locus can be constructed for all annihilation points within the confines of the phantom size. We have adapted this algorithm in MATLAB for 2D PET geometry where we assumed a circular phantom of radius 6 cm and a detector radius of 10 cm to test our model for the reconstruction. A small phantom and geometry were selected to optimize computational efficiency for testing and validating the algorithm. However, a real standard-sized PET was chosen for the later study using the GATE simulated data. We assumed the annihilation photons are detected at A $(0,-10)$ and B $(10,0)$ after one of the photons has suffered single scattering at 90°. For Δl of 0.1 cm, we have a possible set of annihilation points shown with pink and cyan dots in Fig. 2 (left) for the scattering angle of 90°. The annihilation points lie on two circular arcs O1 and O2 in 2D, whose curvature is dependent on TOF. For multiple events with different path length differences (Δl), varying boundary conditions for annihilation are created. Figure 2 (right) illustrates the maximum annihilation area for $\Delta l = 0.1$ cm from all possible scattering angles from 0–90 degrees within the phantom region. Figure 3 shows the reconstruction framework in a flowchart where we change the radius such that the maximum scattering angle is 90°. By superposing all the annihilation areas, we can estimate the location of the annihilation point.

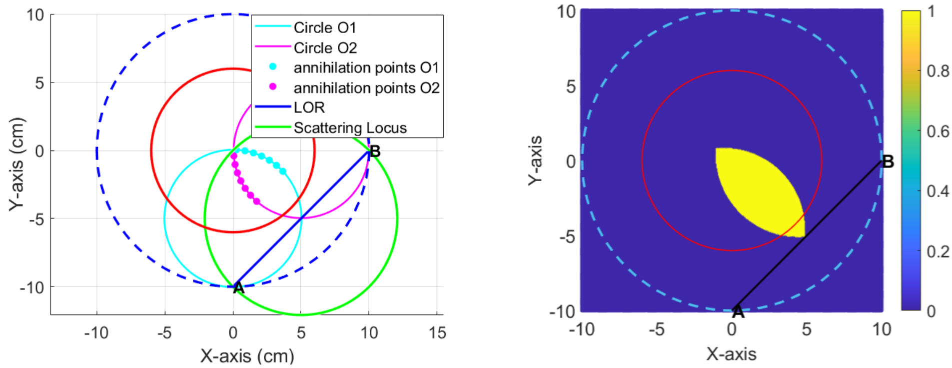

Fig. 2. Left: Loci of annihilation points in cyan and pink for the scattering angle of 90°. Right: A region of annihilation for all scattering angles between 0–90°.

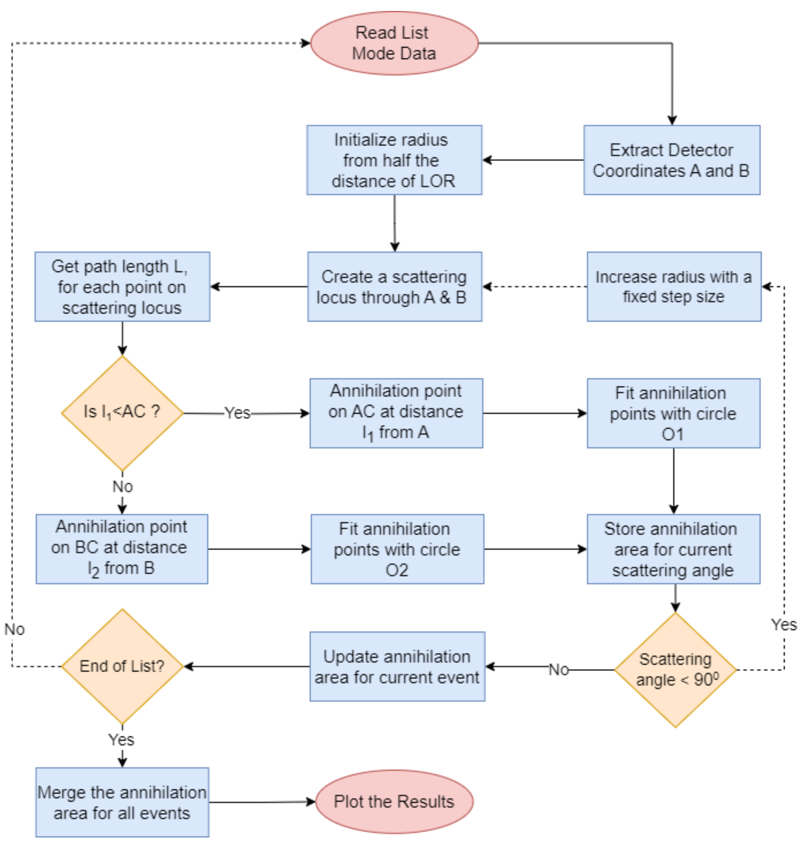

Fig. 3. Flowchart of the reconstruction algorithm for the single-scattered coincidence events.

2.2. GATE simulation of test data

In order to validate the feasibility of the proposed idea, a simulation using GATE [9] was conducted on a benchmark PET system. The simulation involved the use of LSO (Lutetium Oxyorthosilicate) and BGO (Bismuth Germanate) crystals, each having a thickness of 1.5 cm. This algorithm does not take advantage of crystals with higher-energy resolution. We are using the singles event generated from simulation and sorting out the events just based on TOF information. These crystals were arranged back to back, with a shared radius ranging from 46 cm to 49 cm. The detector is partitioned into 8 radial sectors, each measuring 8 cm in width, 32 cm in height, and 40 cm in length. Within each radial sector, there are 16 modules, each subdivided into 25 sub-modules. The configuration of the GATE simulation is shown in Fig. 4.

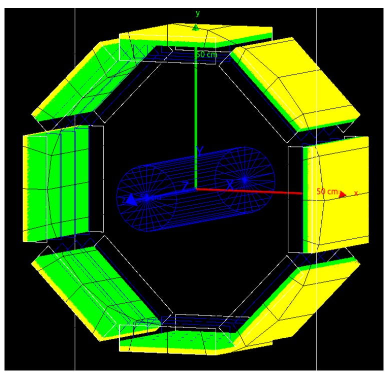

Fig. 4. PET scanner geometry and cylindrical phantom simulated from the GATE benchmark. Yellow is the BGO layer and green is the LSO layer.

We attached a cylindrical phantom of radius 10 cm and height 70 cm filled with water at the center of the geometry with a ^{18}F line source of thickness of 1 mm, situated at the off-center of low activity 1 MBq. The time of hit recorded in the GATE output is the exact time of the event, without any timing uncertainties from the detector's inherent time resolution. After generating the list-mode data from the GATE simulation, we defined a trigger logic to select valid single-scattered events. Events are classified based on the axial difference of less than 1 mm to assume it a 2D geometry, and they turned out to be only roughly 0.01% to 0.02% of the detected events. We created our MATLAB code to filter coincidence events from the singles data created by GATE. We classified our events based on the event ID to make sure the photon pair is from the same annihilation. A coincidence time window of 4–6 ns was chosen for the analysis.

3. Results

The source positioned at 1 cm along both the x and y directions was reconstructed accurately after the trigger application. To enhance image resolution, the reconstructed images were processed through the Lucy–Richardson (LR) deconvolution, which notably improved the clarity of the images. The resolution of the line source obtained was 5.29 cm for direct mapping and 1.79 cm after 50 iterations of the LR deconvolution. In Fig. 5 (left), we present the reconstructed images using single-scattered events with two line sources for 30 s simulation. The first source is kept at $(5, 0)$, while the second one is placed at $(-5, 0)$. Figure 5 (middle) shows an improvement in image quality after the LR deconvolution. The line profile of the deconvolved image is shown in Fig. 5 (right). The model predicts the source location with an accuracy of more than 92%. Application of filters is done to improve the resolution of line source by the algorithm. The accuracy is determined by the error in the peak of the curve from the simulation input.

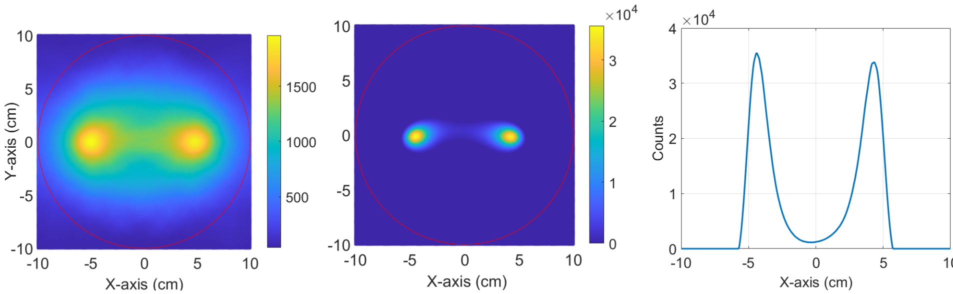

Fig. 5. Left: The reconstructed image for two sources at $(5, 0)$ and $(-5, 0)$ using the single-scatter coincidence events. Middle: Images after 50 iterations of the LR deconvolution. Right: Line profile of the deconvolved image.

4. Conclusion

We introduced a novel algorithm for PET image reconstruction that leverages single-scattered events detected by plastic scintillators using only TOF information. Traditional PET reconstruction methods generally rely on energy information, which is challenging to obtain from plastic scintillators. The proposed method was evaluated using the GATE-simulated data for a 2D PET setup. The results demonstrated that the algorithm effectively reconstructs images from the single-scattered events with the resolution of 5.29 ± 0.12 cm using the point-wise imaging technique. This resolution is improved to the order of 1.79 ± 0.06 cm after the application of the LR deconvolution. A more detailed analysis incorporating realistic timing uncertainties of 200–400 ps is planned for future studies. The ability to use single-scattered data opens up new possibilities for improving PET imaging especially for systems with low coincidence counts. We are currently working on a 3D PET geometry like J-PET to achieve better statistics and that can help improve the spatial resolution and sensitivity of PET scanners.

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