Reconstruction of In-beam PET for Carbon therapy with prior-knowledge of carbon beam-track

Kwangdon Kim[1], Seungbin Bae[2], Kisung Lee[2,3]*, Yonghyun Chung[4], Sujung An[4] and Jinhun Joung[2,3,5]

[1] Department of IT Convergence, Korea University, Seoul, Korea, photon@korea.ac.kr
[2] Department of Bio-convergence Engineering, Korea University, Seoul, Korea, {sbin550,kisung}@korea.ac.kr
[3] School of Biomedical Engineering, Korea University, Seoul, Korea
[4] Department of Radiological Science, Yonsei University, Wonju, Korea, {ychung, crystal8374 }@yonsei.ac.kr
[5] Nucare Medical Systems, Inc., Seoul, Korea, jinhun.joung@nucaremed.com

***** Corresponding Author: Kisung Lee

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** Short Paper*

***Abstract*:** There are two main artifacts in reconstructed images from in-beam positron emission tomography (PET). Unlike generic PET, in-beam PET uses the annihilation photons that occur during heavy ion therapy. Therefore, the geometry of in-beam PET is not a full ring, but a partial ring that has one or two openings around the rings in order for the hadrons to arrive at the tumor without prevention of detector blocks. This causes truncation in the projection data due to an absence of detector modules in the openings. The other is a ring artifact caused by the gaps between detector modules also found in generic PET. To sum up, in-beam PET has two kinds of gap: openings for hadrons, and gaps between the modules. We acquired three types of simulation results from a PET system: full-ring, C-ring and dual head. In this study, we aim to compensate for the artifacts that come from the two types of gap. In the case of truncation, we propose a method that uses prior knowledge of the location where annihilations occur, and we applied the discrete-cosine transform (DCT) gap-filling method proposed by Tuna et al. for inter-detector gap.

***Keywords*:** In-beam PET, DCT gap filling, Hadron therapy

1. Introduction

Positron emission tomography (PET) is known as a useful instrument in cancer diagnosis, localization of suspicious lesions, evaluation of treatment, etc. [1-4].

In cancer treatments, the high-energy bremsstrahlung photons on the MeV scale deliver higher doses to deep-seated tumors while reducing the doses absorbed by the surrounding healthy tissue. Beams of protons and carbon ions have a much more favorable dose–depth distribution than photons [5].

In-beam PET refers to a PET system that can detect the annihilation photons that are generated during hadron therapy. In situ measurement allows one to acquire the maximum statistics by detecting the activity contribution by the very short half-life isotopes, and to minimize blurring effects due to patient shifts during transport to a clinical PET system [6].

However, the geometry of in-beam PET has one or two openings around the rings in order for the hadrons to arrive at the tumor without prevention of detector blocks. The vacancy of detector modules causes truncation in projection data due to insufficient angle coverage [7, 8].

Another problem is an artifact caused by the gaps between detector modules (~6mm). The PET gantry consists of a number of detector modules, and there are gaps between each one. For example, high-resolution research tomograph (HRRT) PET, which is widely known, also has inter-detector gaps due to its eight detector modules. If noise in the acquired projections is very high, as it is with in-beam PET, these gaps may cause artifacts in reconstructed images. It is a kind of research issue to develop algorithms for in-painting, or filling, these gaps without artifacts. One of the simplest algorithms is a bilinear interpolation method that fills the gap with surrounding data in a sinogram with bilinear interpolation. However, the results did not show a perfectly estimated image, and there were secondary artifacts in the recon-

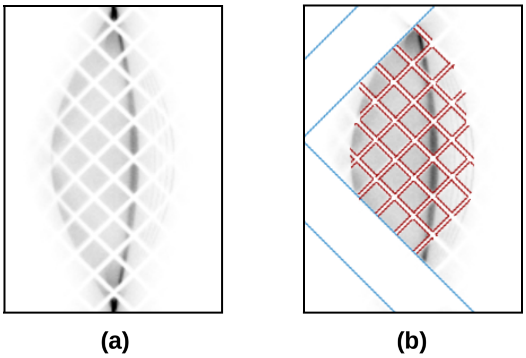

Fig. 1. Sinograms from PET simulation (290AMeV carbon beam): (a) full-ring geometry, and (b) C-ring geometry.

structed image [9]. In transform domain methods, including the constrained Fourier space (CFS) [10] method and the discrete-cosine transform (DCT) domain gap filling method, the algorithm includes a process of applying a mask on, or filtering, the sinogram with a frequency-domain filter.

These methods, inter alia, show superior performance in reconstruction results.

In Fig. 1, part (a) is a sinogram from full-ring geometry simulation data. In the part (b) sinogram, the blue lines represent the gap from limited angle coverage, and red lines represent the gap between detector modules.

In this study, we aim to compensate for the artifacts caused by the blue gaps and to improve reconstruction performance using location information on the pre-determined hadron path. Furthermore, we apply the DCT domain gap filling method on our in-beam PET geometry to suppress the error from the red inter-detector gaps.

2. Materials and Methods

2.1 Simulation

We acquired the list-mode data sets of a PET simulation, where the settings of initial ion energies were 170, 290, 350AMeV of carbon beams modeled with GATE v6.1. Each detector module consists of a 13 × 13 LYSO crystal array. The dimensions of a crystal were 4mm × 4mm × 20mm, and the diameter of the inner circle of the gantry was 30.2cm. In addition, the number of detector modules was 12 to 16, depending on the types of gantry: full ring, C-shaped ring, and dual-head.

We simulated with the full-ring gantry first, as seen in Fig. 2. Then, we took off the events at the detectors located on the hadron beam path that would not really exist in in-beam PET. All simulation conditions are clearly summarized in Table 1.

Table 1. Specifications of simulated PET system.

Parameters	Values
Crystal material	**LYSO**
Crystal size (mm^3)	4 × 4 × 20
Detector module size (mm^3)	52 × 52 ×20
No. of detector modules (full-ring, C-ring, dual head type)	16, 14, 12
Ring diameter (mm)	302
Inter-detector gap length (mm)	6.2

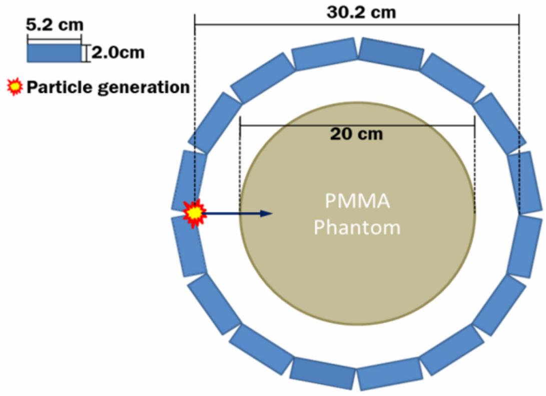

Fig. 2. Geometry of PET detector gantry and hadron beam simulation.

2.2 Reconstruction of Dose Distribution

Low sensitivity is one of the intrinsic problems in in-beam PET because of the lower number of detector modules than with full-ring geometry. The non-uniform placement of detector modules also causes error in the reconstruction image. To reconstruct dose distribution from the high-noise data and asymmetrical geometric data, we developed a reconstruction method based on prior knowledge of hadron beam location.

The basic idea of the proposed method is illustrated in Fig. 3, which shows a diagram of a modified system matrix. To get a modified system matrix, we masked f(x,y) with beam-path information. Then, we calculated H(i,j), the probability of pixel f(i) being detected by LOR p(j). Because most of the annihilation photons are generated along the path of the carbon beam for therapy, we can approximate the origin of photon pairs from the detected line of response, the path of the carbon beam.

A system matrix is a set of probabilities for detecting photons emitted from a certain point in the field of view at a certain point of the detector pixel [12]. Similar to time-of-flight PET reconstruction, which is a method for increasing the spatial resolution of PET by applying a Gaussian distribution to the location of positron annihilation, we applied a Gaussian distribution through the width of the hadron beams. The proposed algorithm subsequently calculates the probabilities under the weighted condition. Then we reconstructed the image iteratively.

2.3 Gap Filling with DCT Domain Filter

An inter-detector gap causes streak artifacts in the background area of a reconstructed image, as shown in Fig. 9 (see arrows). To suppress the artifact, we applied the gap compensation algorithms proposed by Tuna et al. [11].

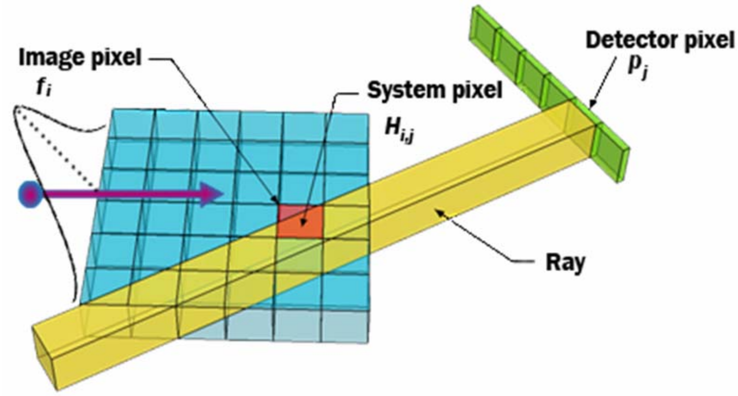

Fig. 3. Diagram for pre-information of annihilating location. The information of the carbon beam is masked on the system matrix.

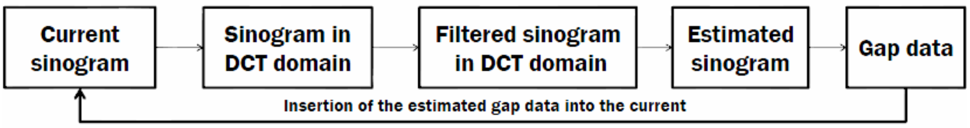

Fig. 4. Flow chart of gap filling method using DCT domain filter mask by Tuna et al. [11].

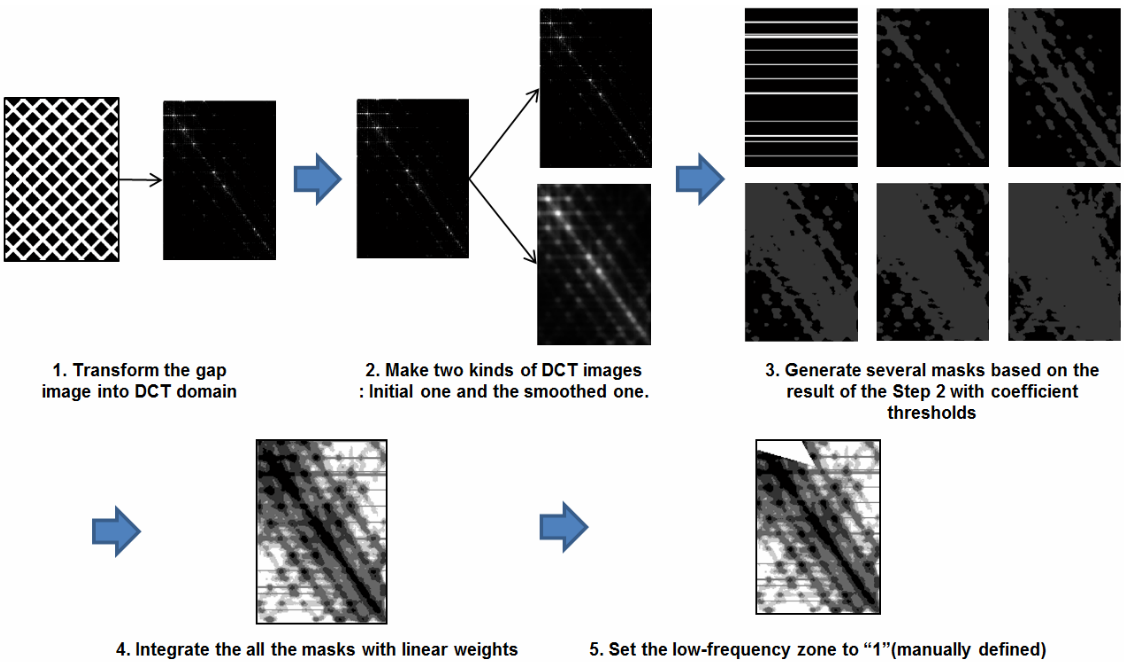

Fig. 5. Flow chart of gap filling method using the DCT domain filter mask of Tuna et al. [11] (DCT domain images in 1 and 2 are contrast-adjusted).

They devised a mask for the gap in the DCT domain for reconstruction in ECAT HRRT PET. The mask is generated by following the steps in Fig. 5, which includes figures for each step in our PET geometry.

Then, the original sinogram in the DCT domain was filtered by the mask via element-by-element multiplication. After the 2-D inverse DCT transformation, an estimate of the full sinogram data is acquired. We extracted the gap area of the full sinogram and overlaid it on the original sinogram. The result replaces the original sinogram, and this series of processes is done iteratively.

2.4 Iterative Reconstruction

Maximum likelihood expectation maximization (MLEM) calculates the most probable activity distribution in the object from the back-to-back gamma ray emission pattern

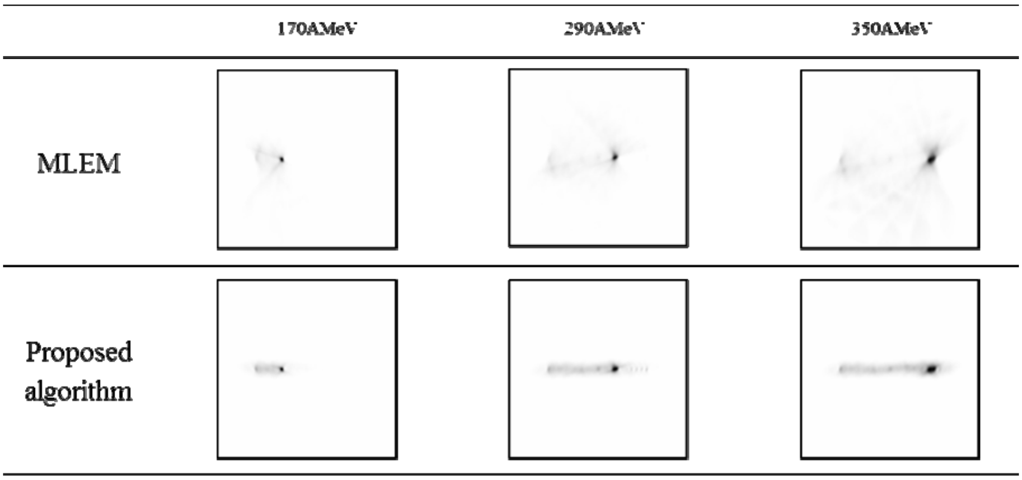

Fig. 6. The results of the proposed method and conventional MLEM classified into energy (C-shaped gantry type).

seen in the PET detection data. The algorithm consists of two steps. The first is forward projection of an assumed activity distribution on the detector and back-projection to recover the image value from projection data [13]:

$$f^{(k+1)}(i) = \frac{f^{(k)}(i)}{\sum H(i,j)} \sum \frac{H(i,j)\,p(j)}{\sum_{i'} H(i',j)\, f^{(k)}(i')} \tag{1}$$

$f^{k}(i)$ denotes the *i*-th image pixel value in the *k*-th iteration, *p(j)* denotes the *j*-th value in the projection data, and *H(i,j)* denotes the element of the system matrix that includes the probability of pixel *f(i)* being detected by LOR *p(j)*.

3. Results

3.1 Prior Knowledge–based Reconstruction

In Fig. 6, we compare the reconstructed dose distributions between conventional ML-EM and the proposed method. The Bragg peak is localized correctly, whereas the surrounding area is represented differently. Images that are reconstructed with the proposed method show not only the Bragg peak point, but the path of the carbon beam.

In Fig. 7, the profile shows the increased gain of data. It seems that low distribution of a conventional iterative PET reconstruction algorithm can be overcome with the proposed method.

In addition, the results of the proposed method show that it can compensate for the error caused by insufficient angle coverage. The profile of the image reconstructed with the proposed method goes through like simulation data, except on the Bragg peak point, which is exaggerated.

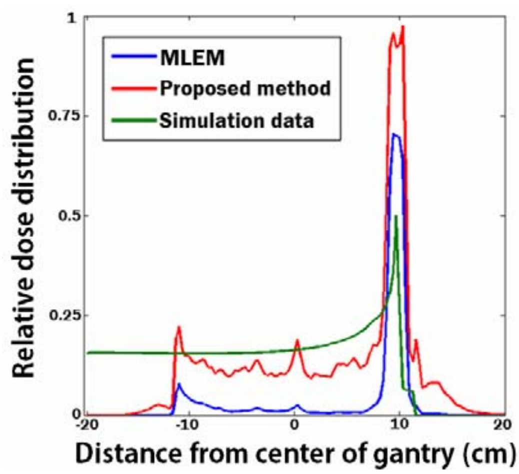

Fig. 7. Normalized profile of the reconstruction image at midline (the initial energy of the beam is 350AMeV).

3.2 Gap Filling with DCT Domain Filter

Fig. 8 shows that the gap filling method reduces the fluctuation values that form streak artifacts in the background. Fig. 9 shows reconstructed images of three different ring geometries, with and without gap filling algorithms. The DCT gap filling method suppresses background artifacts (see arrows) while the open angles of C-shaped and dual head gantries affect image quality significantly.

4. Discussion and Conclusion

In this study, we developed a prior knowledge–based reconstruction and applied a gap-filling method. Through the gap-filling algorithm, the gaps were filled effectively, and we could suppress the artifacts to some extent. We also tried to fill the gap caused by detector module vacancy, but

it did not work well, owing to the large area with this kind of gap. The results of the proposed method can recover the energy distribution through the beam path also observed in fill-ring PET systems.

Beyond the Bragg peak point, there is an overestimated

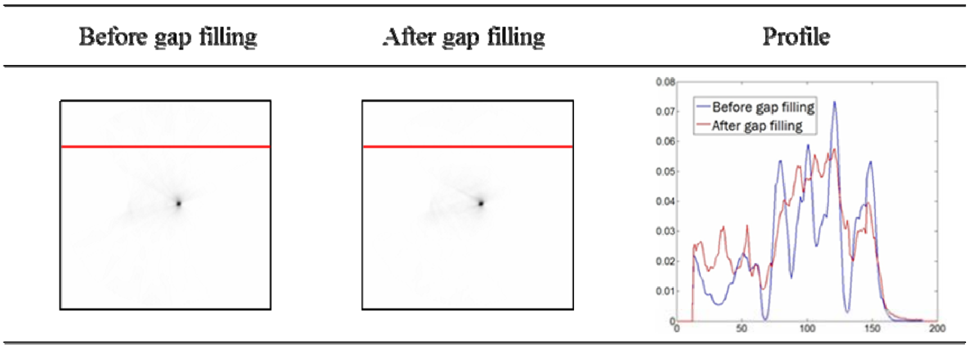

Fig. 8. Profile of red line with and without gap filling image (290AMeV carbon beam; without contrast adjustment).

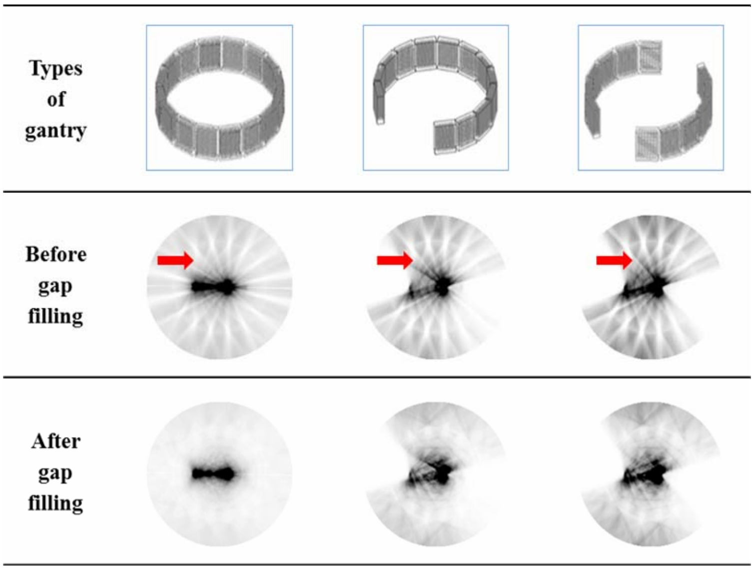

Fig. 9. Comparison between methods, with and without gap filling; 290AMeV carbon beam (contrast is enhanced to identify the streak artifact).

dose value and noise amplification. It should be corrected for more accurate dose distribution. If several differences or artifacts from the proposed method are solved, the method could be another reconstruction solution for in-beam PET. In the final resulting images, the gap-filling method could not provide a conspicuous effect, compared to the prior knowledge-based algorithm, because the insufficient angle coverage has a more significant effect than an inter-detector gap on image reconstruction.

We will improve the way we impose the Gaussian weights through a carbon path to achieve more accurate dose distributions and gap-filling methods to recover an inter-detector gap with more realistic results.

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Kwangdon Kim received his BSc in Radiological Science from Korea University in 2013. Since 2011, he has been a student research member of the Medical Information Processing Laboratory (MIPL) at Korea University. He is now in the MS and Ph.D. course in the Department of IT Convergence, Korea University, Seoul, Korea, and is working in the MIPL.

Kisung Lee received a BSc and an MSc in electronics engineering from Korea University in 1990 and 1992, respectively, and a Ph.D. in electrical engineering from the University of Washington at Seattle in 2003. From 2005 to 2007, he was an assistant professor at Kongju National University, Korea. He has served as a reviewer for IEEE Transactions on Nuclear Science and IET Image Processing. He served as Treasurer of the 2013 IEEE Science Symposium, Medical Imaging Conference and Workshop on Room-Temperature. He is now with the Department of Radiologic Science at Korea University and has worked in the area of radiation detectors, physics in x-ray and gamma-ray imaging systems, and image processing algorithms for medical applications.

Seungbin Bae received his BSc in electronics engineering from Kongju National University, Korea, in 2010. From 2008 to 2010, he was a student research member for NuCare Medical Systems, Inc., Seoul, Korea, working on image reconstruction algorithms for emission computed tomography. Since March 2010, he has worked as a graduate student on detector positioning algorithms for gamma ray detectors and image reconstruction at Korea University, Seoul, Korea.

Yong Hyun Chung has been a Professor in the Department of Radiation Convergence Engineering at Yonsei University since 2006. He received his BSc, MSc, and Ph.D. from the Department of Nuclear and Quantum Engi-neering at KAIST, South Korea, and majored in radiation detection and medical imaging, especially for nuclear medicine instrumentation. From 2002 to 2006, he worked as a Researcher in the Department of Nuclear Medicine, Samsung Medical Center, Seoul, Korea, at the Center for Clinical Research, Samsung Biomedical Research Institute, Seoul, Korea, and at the Crump Institute for Molecular Imaging, David Geffen School of Medicine, UCLA, Los Angeles, USA. His current interests lie in the area of non-invasive imaging techniques, such as gamma cameras, single-photon emission computed tomography (SPECT), positron emission tomography (PET) and specific nuclear material (SNM) monitoring systems.

Su Jung An is a Ph.D. student in the Molecular Imaging Lab in the Department of Radiological Science, Yonsei University, Republic of Korea. She received her BSc in Radiological Science from Yonsei University, Korea, in 2009. She has been working on development of a positron emission tomography (PET) imaging system. As a Ph.D. student, she has been involved in all parts of this research, from detector and data acquisition system design, detector performance measurement and optimization of acquisition parameters for PET imaging.

Jinhun Joung received his MSE and Ph.D. in Biomedical Engineering and Electrical Engineering from the University of Southern California in 1997 and the University of Washington in 2001, respectively. He began his industrial career with the Molecular Imaging Group, Siemens Medical Solutions, USA. While there, he held positions as Principle Research Scientist, Sr. Principle Research Scientist, Patent Manager, and Sr. Project Manager from 2001 to 2009. He served as an elected member of NMISC from 2011 to 2013, and was IEEE NSS/MIC scholarship chair in 2014. Dr. Joung is currently the Chief Executive Officer of NuCare Medical Systems, Inc. and an Adjunct Professor of Biomedical Engineering at Korea University. He has involved himself in various research and development projects in academic and industrial settings related to advanced nuclear medicine detector technology. His research focus is multi-modality molecular imaging detector technology for both in vivo diagnostic and therapeutic imaging procedures.