IV. 5. Development of An Image Reconstruction Method for Planar PEM

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Positron emission mammography (PEM) with $^{18}$F-fluorodeoxyglucose ($^{18}$F-FDG) is a functional imaging technique optimized for breast cancer detection. The development of dedicated imaging systems with high sensitivity and spatial resolution plays a leading role in early breast cancer diagnosis. A prototype PEM consisting of a dual planar head with 4 detector blocks covering field of view (FOV) of about $20\times16\times15$ cm$^3$ was constructed by our group to demonstrate the potential of high-resolution breast cancer imaging in 3-D.

Since most of the well known algorithms for 3-D image reconstruction assumes PET system of cylindrical geometry in order to realize the best performance, it is not simply applicable to nor efficient for planar PEM. The aim of our reconstruction software would be to implement a faster algorithms yet producing fully 3-D resolution which is both homogeneous and isotropic.

Our method is based on ML-EM algorithm which has been widely used for reconstruction of PET images\textsuperscript{1,2}. However, straightforward implementation of ML-EM algorithm for 3-D imaging of our PEM would require a system matrix of Terabytes and a computation time of hours. This is mainly because the number of combination of crystal pairs and voxels is too much for a general purpose computer (PC). Among several solutions commonly known in the case of cylindrical PET, concerning 3-D image reconstruction, we used both sparse property and geometrical symmetry of system matrix to reduce its size and to reduce the computation time.

In planar PEM geometry there exists some symmetries by which we can reduce the size of the system matrix\textsuperscript{3}. Assuming $x$-$y$ plane is parallel to the detector surface, symmetries of $\pi/2$ rotation around $z$-axis and inversion along $x$ - and $y$ - axis tell us that
every line of response (LOR) always finds 7 other LORs (ignoring the symmetrically identical ones) penetrating voxels in exactly the same manner, thus we need only LORs whose polar and azimuth angles are both positive and less than or equal to $\pi/2$. LORs with the same polar and azimuth angles can be found to have the same pattern on penetrating the voxels due to translational symmetry, provided that the crystal pitch is multiple of the voxel width.

Recent PCs are commonly equipped with multi-core processors and/or multiprocessors also capable of vector operation. A parallel computing technique was used to running some calculations at the same time as separate threads, which include a certain LOR and its symmetrical partners. In each thread calculations including at most 4 LORs, which are translationally identical, were also processed at the same time by vector operations such as SSE. The vector-ization were further optimized to use as many 128-bit registers as possible by unrolling the loop and coupling some instructions together in assembler codes so that they would be processed in parallel by pipeline architecture of modern CPU.

OS-EM algorithm is commonly known to be effective in reducing computational costs without loss of image quality$^4)$. It is quite easy to find the best number and order of subsets in the case of cylindrical PET where LORs distribute isotropically. For planar PEM, we found it possible to subsets the projection data according to its parity of index used to identify a crystal pair (indicating $x$ and $y$ coordinates of the crystals) without distorting the reconstruction results. The parity-ordered 4 subsets OS-EM will reduce the number of iteration to 1/4.

The main target of PEM, women breasts are almost homogeneous in its activity and density when seen by annihilation photons and has a relativity simple shape. In particular, it is not difficult to draw a boundary of a volume prior to the reconstruction, which contains most of activities inside and none outside. The fact that in most cases, less than 1/3 of FOV is occupied by such a volume, as shown in Fig. 2 for the case of standard breast size of 565 cc, led us to further reduction of the computation time. In the main loop of the reconstruction algorithm we have incorporated multi-step iterative scheme. In the first step a planar breast image is produced and converted to a 3-D image by simply rotating it, which is used as a initial guess for the next step. Under the ML-EM algorithm, the second and other steps update only a portion of a 3-D image whose voxel value is nonzero, with voxel width gradually decreased beginning from twice to half of the crystal width. The larger voxel size reduces the number of voxels resulting in the faster convergent at the cost of resolution and
which can be fully recovered at the later steps with finer voxels.

The total improvement in computation speed achieved by our method was measured to be faster enough to produce a 3-D image of breast with voxels of 1×1×1 mm³ in less than 1 min.

In order to investigate the quality of the reconstructed image, we performed a series of Monte Carlo simulations using the Geant4 software package⁵, with various scan time and the tumor size and activity concentration. As shown in Fig. 1, we assumed that the random events from outside the FOV can be completely excluded in principle except that coming through the circular window of the shield. The prototype PEM was equipped with arrays of 2×2×15 mm³ detector elements made of a new, fast, high-resolution scintillator, Pr:LuAG⁶. Throughout the simulation study time window was set to 5 ns and a energy window of 445-577 keV was applied, assuming energy resolution of 12% at 662 keV. Small spheres were located along x–, y– and z–axis to simulate breast cancer. Figure 2 shows the typical simulation result in the case of 3 mm diameter spheres with concentration ratio of 10:1 to background activity and a scan time of 10 min. The spheres are clearly seen separated, however, nonuniformity of the image contrast probably due to the absorption effect are also seen.

Both experimental and simulation study are currently under way to incorporate various modifications, such as absorption and scatter correction and random coincidence rejection, into the reconstruction algorithm.

References

Figure 1. Geometry of the PEM detectors and the breast phantom used in the simulation. Pool phantom was used to simulate random events coming from the body of the patient.

Figure 2. Reconstructed image of simulated breast phantom with tumors of 3 mm diameter sphere and a scan time of 10 min.