Monte Carlo Based Reconstruction Using A Rotator for 2-D PET Data

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I. Introduction

Accurate system modeling is normally performed through the System Matrix (SM) in Positron Emission Tomography (PET). The SM can be calculated analytically or by Monte Carlo (MC) methods. MC methods model detector response precisely and lead to better image quality when used in a statistical reconstruction method[1]. However,for modern human PET scanners, two problems hinder the use this method. The first problem lies in the storage of the huge SM. The second problem is the intensive computational load in simulating the SM. We used a Gaussian rotator [2] to solve these problems. With the rotator, both the SM storage and SM simulation requirement is greatly reduced (proportional to the number of scanner blocks) without loss of image quality.

II. Methods and Materials

A. Symmetries

The rotator facilitates the utilization of all the in-plane rotational symmetry of a PET scanner. In Fig. 1, this symmetry is illustrated. The voxel in the rotated grids (in black) contributes the same to the system matrix element as the one in the original grids (in grey). This results in a reduction factor given by the number of detector blocks in contrast to a fixed factor of 8 [3] in a non-rotator case. In 3-D case, other symmetries (i.e. axial translational symmetry [3]) remains.

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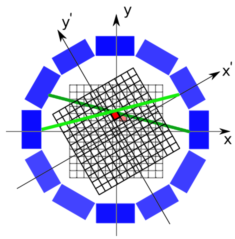

Figure 1. Rotational symmetry. The black grid is the rotated object grid. The grey grid is the original one. The contributions to the green LORs from the red voxels in both grids are identical, if all detector blocks are identical.

B. System Matrix: Calculation & storage

We used the rotational symmetry for the SM calculation and storage. When using the symmetry, any coincidence detection that does not fall into the base SM is rotated to the base SM. In this way the statistics of the SM is improved, thus reducing the total number of coincidences required. In addition, the EGSnrc based code *egs_pet* is used for SM calculations, which has been reported to be at least two orders of magnitude faster than GATE [4]. Only the base symmetry system matrix is stored. A SM element is represented by a 4 bytes unsigned integer, including the SM index and SM value. This leads to a system matrix storage of around $4 \times N_{non\_zero}$ bytes, where N_{non_zero} is the number of non-zero elements in the base symmetry system matrix.

C. Image quality evaluation

To evaluate the image quality, we performed a simulation based study. The image qual-

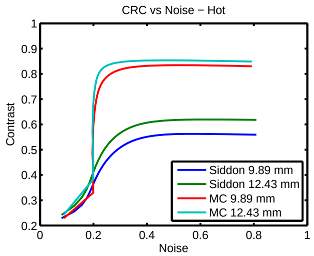

Figure 2. CRC curve for the smallest two hot spots.

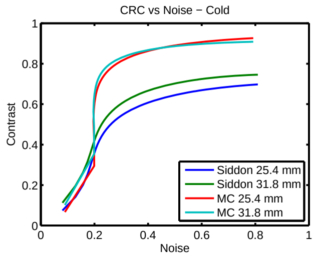

Figure 3. CRC curve for the cold spots.

ity phantom consists of 4 hot lesions (9.89, 12.43, 15.43, 19.79 mm in diameter) with 5:1 ratio to background, 2 cold lesions (25.4, 31.27 mm in diameter) and a background of 204 mm (in diameter). We evaluate the contrast recovery coefficient characteristics of this algorithm. The contrast recovery is measured using the ratio between the mean in the Region Of Interest (ROI) of the lesions and in the ROI of the background. For hot lesions, it is expressed as $C_{contrast} = C \times \overline{H}_{ROI}/\overline{B}_{ROI}$, where C is the true ratio of hot lesions to the background, $\overline{H}_{ROI}$ is the mean in a hot lesion ROI, $\overline{B}_{ROI}$ is the mean in the background ROI. For the cold lesions, it is expressed as: $C_{contrast} = \frac{1-\overline{C}_{ROI}}{\overline{B}_{ROI}}$, where C_{ROI} is the mean in the cold lesion ROI. The noise is evaluated using the standard deviation in the ROI of the background. The results are compared with standard MLEM using a multi-ray ray-tracing (5 rays) projector.

III. Results and Discussions

A. Evaluation results

The CRC (Contrast Recovery Coefficient) curves of 2-D reconstructions are plotted versus noise (Fig. 2 and Fig. 3). The improvement for both hot spots and cold spots is greater than 30%. Moreover, In our method, the CRC reaches to its plateau at lower noise level, especially for hot spots (Fig. 2).

B. Discussion

This rotator based algorithm enables the use of rotational symmetries. The symmetries contribute the a reduction factor of 28 for both storage and simulation time of the SM. The image quality evaluation result indicates that the MC based SM is superior to the ray-tracing system model. However, these improvements are achieved at the cost of significant increase in memory usage and computational time.

IV. Conclusion & Future Work

This proposed methods show superior image quality with regard to standard MLEM. The use of additional rotational symmetries give this method the potential of solving both the SM's storage and calculation problem in 3-D modern human PET reconstruction. Thus, reconstruction based on MC calculated SM for modern human PET systems is achievable. In the future, this algorithm will be extended into fully 3-D reconstruction. Better image qualities can be expected.

References

[1] M. Rafecas, et al, *Use of a Monte Carlo-Based Probability Matrix for 3-D Iterative Reconstruction of MADPET-II Data*, IEEE TNS, vol. 51, no. 5, pp. 2597-2605, 2004

[2] J. W. Wallis, T. R. Miller, *An Optimal Rotator for Iterative Reconstruction,* IEEE TMI, vol. 16, no. 1, pp. 118-123, Feb. 1997

[3] C. A. Johnson, et al, *A System for the 3D Reconstruction of Retracted-Septa PET Data Using the EM Algorithm,* IEEE TNS, vol. 42, no. 4, pp. 1223-1227, Aug. 1995

[4] I. Kawrakow, et al, *Using EGSnrc within GATE to improve the efficiency of positron emission tomography simulations,* AAPM 50th Annual Meeting, 2008