Comparison of 3D SPECT Imaging with a Rotating Slat Collimator and a Parallel Hole Collimator
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Abstract—Besides parallel and converging hole collimators which are frequently used in nuclear medicine, slat collimators can be used for Single Photon Emission Computed Tomography (SPECT). The higher photon collection efficiency, inherent to the geometry of rotating slat collimators results in much lower noise in the data. However, reconstruction of the plane integral data measured by a slat collimator is much more sensitive to noise accumulation compared to traditional SPECT reconstruction from line integrals. It is not a straightforward question whether the initial gain in efficiency will compensate for the larger noise accumulation during reconstruction. Therefore, a comparison of the performance of parallel hole and rotating slat collimation is needed. This study compares SPECT with rotating slat and parallel hole collimation in combination with MLEM reconstruction with accurate system modeling and correction for scatter and attenuation. A contrast-to-noise study revealed an improvement of a factor 3 to 4 for hot lesions and more than a factor of 4 for cold lesion. Rotating slat collimators are thus a valuable alternative for parallel hole collimators.

Index Terms—SPECT, rotating slat.

I. INTRODUCTION

Human SPECT imaging is most often performed using Parallel Hole (PH) collimators. The trade-off between spatial resolution and sensitivity however limits the optimization of SPECT scanners equipped with PH collimators. At the cost of a decreased Field Of View (FOV), converging beam collimators and pinholes offer an enhanced spatial resolution for equal sensitivity and vice versa and therefore are the collimators of choice for organ imaging. Rotating Slat (RS) collimators (figure 1) have the potential for a better trade-off without decreasing the Field Of View (FOV) and can therefore be used as an alternative for PH collimators.

For a parallel hole collimator, the sensitivity is spatially invariant since the decreased photon flux for larger collimator distance \(d\) (photon flux \(\sim \frac{1}{d}\)) is compensated by the larger number of holes seen (holes seen \(\sim d^2\)). Parallel hole sensitivity \(s_{PH}\) is generally described by:

\[
s_{PH} = \frac{g^2}{4\pi h^2}
\]

It can be appreciated that the sensitivity throughout the FOV is only defined by the height of the collimator \(h\) and the hole size \(g\). A more detailed analysis of the sensitivity, taking into account the finite septa thickness and hole shape, can be found in [1]. Since parallel slats only collimate in one direction, a slat collimator is characterized by a different sensitivity function. Because the number of slits seen (slits seen \(\sim d\)) only partly compensates for the decreased photon flux (\(\sim \frac{1}{d^2}\)) at larger distance, the sensitivity of a slat collimated system decreases proportional to the distance to the collimator \(d\):

\[
s_{RS} = \frac{W_0 g}{4\pi h_s (h_s + d_e)} = \frac{W g}{4\pi h (h + d) \cos^3 \alpha}
\]

Furthermore, sensitivity drops with increasing incidence angle due to the smaller detector width 'seen' under an angle \(\alpha\) (\(W = W_0 \cos \alpha\)), the larger effective collimator distance (\(d_e \simeq \frac{d}{\cos \alpha}\)) and the higher effective collimator height (\(h_e = \frac{h}{\cos \alpha}\)), resulting in a \(\cos^3 \alpha\) weighting. A more detailed analysis of the sensitivity and spatial resolution - which is equal for RS and PH - of a rotating slat collimator can be found in [1] and [2]. In figure 2, the sensitivity gain of a RS collimator with respect to a PH collimator in a plane parallel to the slats.

Fig. 1. The geometry used for the calculation of the sensitivity.

Fig. 2. Sensitivity gain of a RS collimator with respect to a PH collimator in a plane parallel to the slats.
$g$ and $h$ respectively are 1.5 mm and 40 mm. The detector width $W$ equals 34.6 cm. By integrating the sensitivity gain profile over the object boundaries for all rotation angles of the camera, one obtains the sensitivity gain for the object of interest. For a cylinder (H=20 cm and radius=10 cm) centered in the FOV and a detector rotation radius of 15 cm, one obtains a sensitivity gain of 36. This gain factor agrees with the findings of Webb [3] who reported a typical sensitivity gain of about a factor 30 to 40.

At first sight, this better photon collection efficiency would lead to better noise characteristics. Unfortunately, plane integral data measured by this device can not be interpreted directly and even for planar images, image reconstruction is needed. For tomographic imaging, all image points in a plane contribute to one measurement in a certain detection bin. The noise in this bin is the added noise over the whole plane and is much higher compared to only the added noise for a line integral in PH collimation. During the backprojection operation of image reconstruction, the higher noise is backprojected to a lot more image points in the case of RS, which finally results in higher noise accumulation. Figure 3(a) illustrates the effect of the higher noise accumulation by starting from an equal number of counts in the measured data for both a PH and a RS collimator. This mimics the hypothetical case of no gain in geometric sensitivity for the RS collimator. It can be clearly seen that the RS image is deteriorated a lot more by noise compared to the PH image when both images are converged to equal contrast. The noise accumulation induced by the reconstruction compromises the gain in geometric efficiency and thus leads to a lower effective efficiency after reconstruction.

Planar RS imaging with Filtered Back-Projection (FBP) reconstruction has previously been studied by Lodge et al. [4]. Their work shows an advantage in SNR over a PH collimator for small activity distributions and enhanced contrast for small hot spots. A comparison by our group uses accurate system modeling in an iterative reconstruction and indicates improved contrast-to-noise ratios up to a factor 3, even in large objects approaching the size of the FOV of the camera [5].

In the tomographic case, FBP reconstructed SPECT images with a RS collimator again confirmed improved noise characteristics for small objects (smaller than 10 cm) and showed improved contrast only for small regions of high tracer uptake [6]. An iterative reconstruction approach found better contrast-to-noise for the RS for both cold and hot lesions [7]. However, in this study a solid state strip detector [8] was used and the image quality improvement was not only due to collimation with slats but was also subject to the combined effect of small detector width, better collimator resolution and solid state detector. Moreover, in this comparison, no model for depth dependent detector blurring was used during reconstruction.

To investigate the impact on image quality of only the RS collimator in combination with iterative reconstruction, a fair comparison is needed. This study compares fully 3D SPECT with RS and PH collimation in combination with MLEM reconstruction with scatter compensation, attenuation correction and accurate system modeling.

II. METHODS

A. Phantom simulation

The Monte Carlo simulated data used in this study are generated using GATE [9]. The image quality phantom (Standard Jaszczak Phantom, shown in figure 4 is simulated containing 4 hot spheres (diameters: 9.9 mm, 12.4 mm, 15.4 mm, 19.8 mm) and two cold spheres (diameters: 24.8 mm and 31.3 mm). The activity concentration is set in order to have a sphere-to-background activity ratio of 8:1 in the hot spheres. The total activity in the phantom was 37.6 MBq.

The phantom is simulated to be filled with water with an attenuation coefficient $\mu$ of 0.154 cm$^{-1}$ at 140 keV. Figure 4(a) shows the three headed rotating slat camera that was modeled while in figure 4(b) the three headed parallel hole system is shown. Both the PH and the RS camera consist of an identical detector modeled as 192x192 individual pixels of 1.8x1.8 mm. 193 parallel slats of height 40 mm and 0.3 mm thickness model the RS collimator. This design is based on the SOLSTICE design proposed by Gagnon [8] but uses a complete detector instead of only a strip of detection material. The PH collimator is implemented as a square hole collimator with height 40 mm and 0.3 mm septal thickness. The hole pitch is 1.8 mm and matches the slat pitch of the RS collimator. The collimator resolution is thus matched for both collimators and is 5 mm at 10 cm collimator distance. The rotation radius of the detector was 15 cm and the acquisition time was set to 8 minutes. 10 realizations of the same acquisition were...
simulated. For all events recorded, a flag was enabled in case a Compton scatter interaction occurred in the water phantom.

B. Scatter correction

In the GATE simulations, a realistic phantom was modeled and consequently the datasets are contaminated by scatter. In order to correct for scatter in the data, the Dual Energy Window (DEW) scatter estimation technique [10] was used to correct the data acquired with a rotating slit collimator. The scatter corrected data is calculated using the DEW technique as follows:

\[ g_{SC} = g_{MW} - k \left( \frac{w_{MW}}{w_{SW}} \right) \]  

(3)

with \( g_{SC} \), \( g_{MW} \) and \( g_{SW} \) respectively the scatter corrected data, the data measured in the main energy window and the data measured in the scatter window. The width of our main energy window \( w_{MW} \) was chosen 14 keV around 140keV and \( w_{SW} \), the width of the scatter window was chosen 10 keV and located around 125 keV. Both energy windows are shown in figure 5 where also the estimate of the scatter with the above formula is given for \( k = 0.5 \). Next, also the scatter corrected data measured in the scatter window. The width of our main energy window \( w_{MW} \) was chosen 14 keV around 140keV and \( w_{SW} \), the width of the scatter window was chosen 10 keV and located around 125 keV. Both energy windows are shown in figure 5 where also the estimate of the scatter with the above formula is given for \( k = 0.5 \). Next, also the scatter corrected data measured in the scatter window.

true scatter is plotted and it can be seen from the energy spectrum of the scatters that there is indeed a good agreement between the estimate and the true scattered photons, which were flagged during simulation. For validation, the number of counts, estimated with the DEW technique and the true number of scattered detections were compared for the ten realizations of the data using the mean absolute error (MAE). The MAE between estimated and true number of scatters over the different realizations was 0.6%. For comparison, also the MAE for the PH data was calculated resulting in a value of 1.4%. This indicates the DEW scatter correction method is valid for data collected with a RS collimator, although no attempt is made to make a spatial validation at the level of the projections.

C. Image reconstruction

For the purpose of reconstruction of the plane integral data collected by the RS collimator, we used a previously proposed split-matrix method for accelerated image reconstruction [11], [12]. This method does not model the process to go from plane integrals to image in one single step, but splits the process in two separate steps, the first is a model \( A \) to go from plane integral data to sinograms and the second models the step to go from sinograms to image \( B \). The resulting system matrix \( AB \) is used in an MLEM algorithm:

\[ f_k^{t+1} = \sum_i B_{ki} \sum_j A_{ij} \sum_k A_{kj} g_j \sum_i A_{ik} \sum_k B_{ki} f_k^t. \]

The following iterative process is followed by this algorithm: in a first step, an initial image estimate \( f_k^0 \) is forward projected with system matrix \( B \) to a sinogram estimate. Next, this sinogram estimate is forward projected using \( A \) and compared with the scatter corrected plane integral data \( g_j \). The resulting plane integral update is projected backward using \( A \) and immediately projected backward with \( B \) to image space where it serves as an update image. After updating and normalizing the original image estimate, the next iteration can start.

In the first part of the system matrix \( B \), which transforms image space to sinogram space, we model the depth dependent sensitivity as the mean sensitivity in a plane parallel to the detector. Furthermore, depth dependent resolution is also modeled in \( B \). System matrix \( A \) involves a mapping from sinogram to plane integral space. At this point, we do not include any sensitivity modeling nor resolution modeling.

D. Attenuation correction

1) Implementation: Since the data are affected by attenuation, we need a method to compensate for the loss of photons at larger depths. An attenuation correction method which fully models the attenuation along each possible ray of projection as used by Zeng et al. [13] is used for the reconstruction of the PH data. However, for RS data, this method is only possible in a fully 3D reconstruction algorithm which directly maps image to the plane integrals using one single system matrix. Since our fast reconstruction method splits the system matrix in two separate ones, the original paths of detection are lost and this method can not be applied. Therefore, we base ourselves on Chang’s attenuation correction method [14] which calculates an average attenuation factor at each voxel. However, where Chang calculates the mean attenuation value over all SPECT angles for every voxel, our method will calculate an average attenuation coefficient \( c \) over all spin angles \( \phi \) for every voxel \( (x, y, z) \) and for every SPECT angle \( \theta \):

\[ c(x, y, z, \theta) = \frac{1}{\phi} \sum_{i=0}^{\phi} \exp\left(-\frac{(ML)_{\phi_i}}{\phi}\right). \]

(4)

with \((ML)_{\phi_i}\) the mean attenuation length product for spin angle \( \phi_i \). The calculation of \((ML)_{\phi_i}\) was done by tracing 100 paths connecting the voxel of interest with equidistant points, sampling the detector. In figure 6 (a) one of such paths is displayed. The ray tracing, using Siddon’s algorithm [15].
returned the intersection lengths of these rays with the attenuation image voxels. Sensitivity weighting of the intersection lengths with \( \cos^3 \alpha \) yielded a map of weighted intersection lengths for a certain voxel (figure 6 (b)). Finally, multiplication with the \( \mu \) value of the intersected voxel and averaging over all possible paths resulted in \((ML)_{ij}\).

Since our system matrix \( B \) rotates (Gaussian rotator) the image according to the appropriate SPECT angle before applying sensitivity and resolution modeling, we can use the factors \( c \) to compensate the rotated image for attenuation. In this way, a semi-average attenuation compensation is performed in system matrix \( B \), being calculated with every SPECT angles as an average over all spin angles.

2) Validation: GATE was used to validate the correctness of the attenuation values \( c \) of our analytic calculation method. Four high count simulations (noise<0.8%) were performed, resulting in four datasets \( I_x, I_x,0, I_y, I_y,0 \). For the generation of \( I_x \) and \( I_y \), line sources aligned with respectively the X and Y axis of the scanner (Fig. 1) were simulated with a cylindrical water phantom present. The camera was spun over 360 degrees. Binning the recorded data according to the source position resulted in the datasets \( I_x, I_x,0, I_y, I_y,0 \) were generated in a similar way with the only difference of the absence of the water cylinder. Since the detector was rotated over all spin angles there exists the following relation between datasets \( I \) and \( I_0 \):

\[
I = I_0 \frac{1}{\phi} \sum_{\theta=0}^{\phi} \exp(-(ML)_{ij}),
\]

(5)

\( I/I_0 \) should thus be equal to the reciprocal of the attenuation values \( c \). The calculations of \( c \) were based on an attenuation map derived from the cylindrical phantom. In figure 7 the results of the simulation and calculation are summarized as a line profile through the reciprocal of the calculated attenuation values \( c \) for respectively \( y = 0 \) and \( x = 0 \) in figures 7 (a) and (b) and as the ratio of respectively \( I_x, I_x,0, I_y \) and \( I_y, I_y,0 \) in figures 7 (a) and (b). These profiles show very good agreement and suggest the proposed analytic calculation method for the attenuation values is suitable for attenuation compensation during our iterative reconstruction. Transaxial slices through reconstructed images of a uniform cylindrical source are shown in figure 8 (a) and (b) in the case of respectively no attenuation correction and attenuation correction with the calculated correction factors \( c \).

III. RESULTS

A. Contrast-to-noise

Figure 9 shows the contrast-to-noise plot for the 19.8 mm hot spot. The hot spot shows a better contrast recovery for RS compared to the PH collimator. To have an idea about the quantitative improvement, the imaging time was increased for the PH collimator. The results for a 2, 3 and 4 times longer PH acquisition are also plotted in figure 9. From these plots, it can be seen that, to obtain the same contrast-to-noise, the PH collimator needs about a 3 to 4 times longer measurement compared RS collimator. The results for the 31.3 mm cold spot are shown in figure 10. This plot shows that the RS collimator needs less than 4 times the imaging.
time of a PH collimator in order to obtain similar contrast-to-noise characteristics. Compared at an equal noise level of 50%, contrast recovery is 28% higher for the hot lesion while it is 30% higher for the cold lesion. For the RS collimator 40 iterations are needed to obtain 50% noise while the PH only needs 10 iterations to achieve this noise level.

B. Tomographic images

A transverse, coronal and sagittal slice through the images at equal noise (40%) is shown in figure 11 and in figure 12 for respectively the PH and the RS collimator. It can be seen from the images already that the RS collimator improves the contrast. The profile drawn through the lesions in figure 13 confirms the better contrast recovery with the RS collimator.

IV. DISCUSSION AND CONCLUSION

Scatter correction was used and found at least equally accurate compared to DEW scatter correction for PH collimated acquisitions. Furthermore, an analytic method for calculating attenuation factors for every voxel at every SPECT angle was proposed and validated. This enabled us to correct for attenuation in a previously proposed reconstruction method for plane integral data. With these tools available, a realistic Monte Carlo simulation of an image quality phantom could be reconstructed. The results show that SPECT images obtained with a rotating slat collimator in combination with iterative reconstruction and accurate modeling provide a much better contrast-to-noise trade-off compared to images obtained with an equivalent PH collimator. This study not only shows better hot spot contrast recovery, which was also found by Lodge, but also proves better cold spot contrast recovery. This result
already had been obtained by Wang with a slat collimated solid-state strip detector, but in that study, no depth dependent resolution was modeled. Also, it was unclear whether the improved contrast-to-noise was due to the better solid state detector or due to the slat collimator. Recently, it was shown by Defrise in a technical note [16] that the optimal collimator resolution for a RS collimator has to be better by a factor $\sqrt{2}$ relative to the optimal PH collimator resolution. These findings could be in favor of the RS collimator and should be investigated in more detail in the future.

The better cold spot contrast recovery confirms that in the tomographic case, a PH collimator also suffers from noise accumulation during reconstruction. This concludes that rotating slat collimators can be a valuable alternative for parallel hole collimators, with shortening of the imaging time by a factor of 3 to 4.

REFERENCES