Abstract—Transrectal Compton imaging probe is capable of simultaneous high sensitivity and high resolution that are necessary for detecting lesions with small sizes. However, it suffers from the limited angle tomographic reconstruction. Consequently, the quality of the reconstructed images is significantly degraded. We present a practical and effective scheme to eliminate limited angle problem inherent to the proposed Compton imaging probe by using auxiliary data to augment the incomplete tomographic data acquired from the probe alone. We give a mathematical formulation for the proposed scheme and describe its implementation. The proposed scheme fits seamlessly into the framework of the list-mode maximum likelihood expectation maximization reconstruction algorithm, yielding superior reconstructions. Our preliminary results demonstrate the effectiveness of the proposed approach.

Index Terms—Compton imaging probe, Compton camera, limited angle tomography.

I. INTRODUCTION

The wide awareness and practice of early detection have led to a drastic downward stage migration in prostate cancer. Consequently, the size of lesions to be detected has been continuously decreasing. The detection of the small lesions pose an urgent demand for an imaging system with high spatial resolution and high detection sensitivity. In [4], we have proposed a gamma-ray imaging probe that is capable of simultaneous high sensitivity and high resolution based on Compton scatter camera techniques for prostate imaging. However, such a device suffers from the limited angle problem because the probe alone only views the prostate over a restricted angle, resulting in degradation in the quality of reconstructed images in several aspects. Most importantly, the emission position of the source cannot be fully determined, which is particularly problematic with the existence of high non-prostate activity. Counts collected due to the non-prostate radiation contribute a significant background in the reconstructed images. This is further exacerbated by reconstructing an image volume that is smaller than the volume containing source activity, when some of the source activity outside the reconstructed volume is seen by the detector. Furthermore, like all limited angle tomography, spatial resolution in the direction perpendicular to the detector is limited. In this paper, we aim to alleviate the image degradation associated with the limited angle tomographic reconstruction of Compton probe data. Specifically, the measurement due to the non-prostate activity is modeled as additive Poisson random variables in the likelihood function. SPECT projection data acquired over a complete angular range for the same source distribution as that for the probe is used to estimate the background in the prostate probe projection data.

II. METHOD AND ALGORITHM

A. System description

We chose a design previously proposed in [4] as our system in the subsequent investigation. Figure 1 shows the configuration of the system. It comprises a transrectal probe, which consists of a stack of 1 mm thick 1 cm × 4 cm pixelated silicon detectors that are placed intrarectally and operate in conjunction with an external NaI(Tl) gamma camera in time coincidence. The pixel size of the silicon detector is 1 mm × 1 mm. The gamma camera comprises two 40 cm wide, 40 cm deep and 2 cm thick NaI(Tl) crystals with a 3 mm FWHM intrinsic spatial resolution, and is placed above and below the pelvic region and close to the patient to maximize the solid angle subtended by the camera to probe. This configuration is particularly suited to imaging low energy gamma-ray radiotracers. Specifically, good spatial resolution can be attained due to the extremely close distance between the probe and the prostate (1 cm).

![System diagram](image)

Fig. 1. System diagram.

B. Methodology

This section describes the reconstruction scheme that is effective to reduce the limited angle problem inherent to the Compton imaging probe. The proposed scheme fits seamlessly into the framework of the list-mode ML-EM algorithm and uses auxiliary data to augment the incomplete Compton probe data. Specifically, a complete tomographic data set is formed by integrating the high resolution, but limited angle probe data with low resolution, yet complete SPECT or PET data. In practice, a complete data set can be derived from either a conventional SPECT or PET scan. Although low resolution SPECT/PET data are limited in providing detailed information about the prostate, they cover a big field of view (FOV) over a complete angular range and therefore provide full information on non-prostate background that can be used to eliminate the noise.
contributed by background sources. The images reconstructed using the combined multi-resolution data in the framework of the ML-EM are expected to retain high spatial resolution (offered by the probe data), and exhibit an improved contrast as a result of background reduction. One significant challenge of such an approach is prohibitively high computation complexity. Because a full FOV is to be reconstructed instead of a small window, computation is daunting even with a coarse pixelation in the non-prostate regions. One alternative to the aforementioned scheme is to restrict the reconstructed volume over the prostate region only, but correct the background measured in each projection acquired by the Compton probe based on the existing SPECT data. This is explained with the illustration of Figure 2. The box in Figure 2 depicts the reconstructed volume that encloses the prostate. Each Compton probe projection is corrupted with background contributions from non-prostate sources, which can be estimated by integrating the activity density outside the prostate along the backprojected cone for each measured projection.

To reduce the reconstruction time, we propose an alternative approach. Specifically, in our scheme, the mean activity in non-prostate regions is assumed a known quantity, resulting in

$$\lambda_{j}^{(n+1)} = \frac{\lambda_{j}^{(n)} + \sum_{i} a_{pi} Y_{pi} + \sum_{j} a_{cij} Y_{cij}^{(n)}}{\sum_{i} a_{pi} + \sum_{j} a_{cij} \lambda_{j}^{(n)} + \beta r_{ij}}$$

where $b_{pi}$ represents background measured in the $i$th projection $Y_{pi}$. Consequently, the update in the M-step of the penalized ML-EM algorithm becomes

$$\lambda_{j}^{(n+1)} = -\frac{1}{2} \left( \frac{1}{\beta r_{ij}} \right) + \frac{1}{4} \left( \frac{1}{\beta r_{ij}} \right)^{2} + \frac{1}{2} \lambda_{j}^{(n)} \sum_{i} a_{ij} \lambda_{j}^{(n)} + b_{ij}.$$  

Note that the number of parameters needs to be estimated in our scheme is significantly reduced as the mean activity outside of the limited reconstruction volume is assumed a known quantity. As a result, this approach greatly reduces the computational complexity. Furthermore, the discrepancy between the reconstructed image generated by our scheme and the one derived by the full-object construction is expected to be small due to the considerably poor spatial resolution of the complete SPECT data.

**D. SPECT and Compton probe data generation**

We have tested the proposed method using simulated SPECT and probe data. In our experiment, a SPECT camera with a NaI crystal of 50 cm × 50 cm × 0.95 cm in conjunction with a medium energy, high resolution collimator was simulated to generate the point spread functions (PSFs) for $^{111}$In. The NaI detector has a FWHM intrinsic spatial resolution of 4.6 mm and energy resolution of 10% for both emissions of $^{111}$In. The simulated collimator has hexagonal holes with a hole diameter (distance between opposite faces) of 2.07 mm, septal thickness of 0.68 mm and hole length of 32 mm [5]. The sensitivity of the collimator is 11 cpm kBq$^{-1}$ for $^{111}$In. Radial PSFs at 0 cm, 10 cm, 20 cm, and 30 cm from the front face of the collimator were simulated separately for the 171 keV and 245 keV photons emitted from a $^{111}$In point source in the Zalib phantom. At 171 keV and 245 keV, the septal penetration and scatter are negligible. To properly model the penetration, each simulated PSF was fitted using a Gaussian plus an exponential function as given by,

$$PSF_{d_{i}}(r) = A_{\text{Gauss},d_{i}} \exp \left( \frac{-r^{2}}{2\sigma_{d_{i}}^{2}} \right) + A_{\text{exp},d_{i}} \exp \left( -\lambda_{d_{i}} r \right),$$

where $d_{i}$ represents the distance from the source to the collimator for the $i$th simulation. Figure 3 shows the horizontal and vertical profiles of the simulated PSFs at four simulated distances. The solid lines denote the fitting functions using equation (5) as the template. The figures on the top row represent the PSFs at 171 keV and the figures on the bottom row show the PSFs at 245 keV. A good agreement between the
1.0 for the 28 94 σ 0.4 a use A λ 1 111 were then used to derive a set of (1002 1874, 0.6 σ A 2 2.0). 0, 1.0 λ (AND λ 1 488 × 201 A 111 at each energy, ∑ −1 use a summation of two exponential functions as the −2 1.5 (56 d σ and 1.0 λ, phantom with a typical . Figure 4 (b) shows the image of the Relative Activity Con-

TABLE I
RELATIVE ACTIVITY CONCENTRATIONS IN VARIOUS ORGANS IDENTIFIED AS MAJOR SOURCES OF BACKGROUND AT DAY-FIVE OF CLINICAL 111In PROSTA SCINT SCANS.

<table>
<thead>
<tr>
<th>Organ</th>
<th>Relative Activity Concentration</th>
<th>Volume (ml)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Prostate</td>
<td>1.0</td>
<td>28</td>
</tr>
<tr>
<td>Liver</td>
<td>2.0</td>
<td>1874</td>
</tr>
<tr>
<td>Blood</td>
<td>1.5</td>
<td>481</td>
</tr>
<tr>
<td>Bone marrow</td>
<td>0.7</td>
<td>1002</td>
</tr>
<tr>
<td>Kidney</td>
<td>1.0</td>
<td>488</td>
</tr>
<tr>
<td>Spleen</td>
<td>1.0</td>
<td>356</td>
</tr>
<tr>
<td>Bladder</td>
<td>0.6</td>
<td>201</td>
</tr>
<tr>
<td>Rectum</td>
<td>0.4</td>
<td>94</td>
</tr>
<tr>
<td>Testes</td>
<td>0.6</td>
<td>56</td>
</tr>
</tbody>
</table>

Simulated PSFs and parameterized PSFs derived by the curve fitting is shown for all distances and for both energies. The PSFs at 245 keV have a wider tail in comparison to that at 171 keV due to the increased septal penetration and scattering in the collimator, resulting in a flatter exponential slope in the fitting functions. The resulting coefficients $A_{Gauss,d_i}, \sigma_{d_i}, A_{exp,d_i}, \lambda_{d_i}$ of PSF at each distance $d_i$ were then used to derive a set of functions that compute coefficients $A_{Gauss}, \sigma, A_{exp}, \lambda$ for the PSF at each energy, using the source-to-collimator distance. Specifically, the functions for the computation of $A_{Gauss}$ and $A_{exp}$ use a summation of two exponential functions as the template. The function for the computation of $\sigma$ and $\lambda$ use the third order polynomial function as the template. With the functions that can generate $A_{Gauss}, \sigma, A_{exp}, \lambda$ at each energy, depth-dependent PSFs can be derived at various source-to-collimator distances and both energies.

The true object, namely the Zubal phantom with a typical 111In ProstaScint activity distribution shown in Table I, serves as the initial images. The forward projection operator in the software package ASPIRE [1] was used to produce noiseless projection data for each energy based on the corresponding PSFs and known attenuation maps. The resulting projection data at each energy were scaled to an equivalent imaging time of 60 min as in clinical SPECT scans. The projection data within the two energy windows were then summed to yield the noise free projection data for 111In. Poisson noise was next added to the noiseless projection data to generate the realistic clinical SPECT projection data. A 128×128×128 3-D SPECT image with the isotropic voxel size of 4 mm$^3$ was reconstructed from the simulated projection data. Filtered backprojection using Chang’s attenuation correction scheme was applied [3]. The Hanning window with a cutoff frequency of 0.9 cycles/pixel was used. A Monte Carlo simulation code developed based on EGS4 was used to generate data for the transrectal Compton imaging probe. The silicon detector and dual-head gamma camera described in Subsection II-A were used. The silicon detector was assumed having 1 keV energy resolution. The NaI detectors have a FWHM intrinsic spatial resolution of 3 mm and an energy resolution of 10%. In the probe simulation, the phantom and background distribution were same as that in SPECT studies. An 8 mm tumor with a T/B ratio of 10:1 was located inside the prostate. For each individual detection events, we then estimate the mean contribution of the background, based on the reconstructed SPECT images. Limited by the available memory, the reconstructed SPECT images were downsampled by a ratio of 2:1. The background measurement was estimated by projecting the activity outside the prostate in the reconstructed SPECT images along the backprojected cone defined by each probe measurement. Note that proper scaling is needed to account for both the discrepancy between the voxel sizes of the SPECT and probe, and the differences in the efficiencies of two imaging systems. A total of 1.3M counts were collected and three equi-spaced tomographic slices were reconstructed. Of them, one slice was within the prostate and was located 5 mm from the center of the simulated tumor.

III. RESULTS AND DISCUSSION

Figure 4 (a) shows the reconstructed image of the prostate by ignoring the background measured in the projection data of the Compton probe using ML-EM algorithm at 100 iterations. Sever artifacts are clearly seen around the corners of the images. This is because projections stemming from the source lying outside the reconstructed volume is amplified due to a small value of $\sum_j a_{pi}/\lambda^{(n)}_j$. Figure 4 (b) shows the image of the prostate using update equation (3) with the same number of iterations. The artifacts in Figure 4 (a) have been eliminated. In addition, the image has less background compared to that in Figure 4 (a). Figure 5 represents superimposed horizontal profiles through the tumor center for both images. The profiles are normalized so that their maximum intensity is same. The proposed method yields better tumor-to-background contrast. In fact, modeling the background contribution in each projection results in slower convergence. Therefore, images with better quality can be generated if more iterations are performed in our scheme.

![Fig. 5. Superimposed profiles of both images through the tumor center.](image)

IV. CONCLUSIONS

We propose a practical scheme to eliminate or alleviate limited angle tomographic reconstruction for the Compton
imaging probe. The proposed scheme has been seamlessly integrated into the framework of the list-mode ML-EM algorithm. Our preliminary results demonstrate the effectiveness of the proposed approach.

It is worth mentioning that one can employ a dual-probe geometry, in which an additional external probe is placed beneath the perineum, perpendicular with respect to the transrectal probe as shown in Figure 6. Such a configuration results in high spatial resolution data in both transverse and depth direction. Previous studies have shown that a significant improvement is achievable with the use of an orthogonal view in the limited angle tomography [2]. This approach is expected to improve the spatial resolution of the transrectal probe along the direction that is perpendicular to the transrectal probe. The effectiveness of this approach needs to be investigated. Our future work includes the full scale reconstruction of the probe data augmented with the auxiliary SPECT data and the combined reconstruction of the transrectal and external probe data.

REFERENCES


Fig. 3. The vertical and horizontal profiles of the PSFs of the simulated SPECT system for 171 keV and 245 keV gamma-rays of $^{111}$In at 0 cm, 10 cm, 20 cm, and 30 cm from the collimator. Symbols denote Monte Carlo simulated results and solid lines represent fitted results.

Fig. 4. A slice of the reconstructed prostate images. (a) without estimating the background in each projection. (b) estimating the background in each projection.

Fig. 6. Enlarge limited angle tomographic data using two Compton probes.