A novel three-dimensional image reconstruction method for near-field coded aperture single photon emission computerized tomography

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Coded aperture imaging for two-dimensional (2D) planar objects has been investigated extensively in the past, whereas little success has been achieved in imaging 3D objects using this technique. In this article, the authors present a novel method of 3D single photon emission computerized tomography (SPECT) reconstruction for near-field coded aperture imaging. Multiangular coded aperture projections are acquired and a stack of 2D images is reconstructed separately from each of the projections. Secondary projections are subsequently generated from the reconstructed image stacks based on the geometry of parallel-hole collimation and the variable magnification of near-field coded aperture imaging. Sinograms of cross-sectional slices of 3D objects are assembled from the secondary projections, and the ordered subset expectation and maximization algorithm is employed to reconstruct the cross-sectional image slices from the sinograms. Experiments were conducted using a customized capillary tube phantom and a micro hot rod phantom. Imaged at approximately 50 cm from the detector, hot rods in the phantom with diameters as small as 2.4 mm could be discerned in the reconstructed SPECT images. These results have demonstrated the feasibility of the authors' 3D coded aperture image reconstruction algorithm for SPECT, representing an important step in their effort to develop a high sensitivity and high resolution SPECT imaging system. © 2009 American Association of Physicists in Medicine. [DOI: [10.1118/1.3103490](https://doi.org/10.1118/1.3103490)]

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I. INTRODUCTION

Single photon emission computerized tomography (SPECT) is a noninvasive imaging technique that has been widely used in the detection of myocardial perfusion abnormality and the evaluation of left ventricular function for more than three decades. With the recent improvements of gamma detectors and image reconstruction algorithms, this imaging modality has the potential of imaging the molecular processes, such as angiogenesis and apoptosis in the ischemic myocardium.¹ These kinetic processes can be quantified *in vivo* from molecular-targeted SPECT images,²⁻⁴ providing in-depth knowledge which may lead to the future development of radioactive and pharmaceutical agents for clinical diagnosis and therapy for heart disease. Compared with positron emission tomography (PET), another molecular-targeted imaging technique, SPECT has several advantages including the relative ease in labeling endogenous ligands due to greater tracer availability, the capability of measuring slow kinetic processes due to the long half-life of commonly used radioactive tracers, and the ability to image multiple molecular pathways with simultaneous multiple energy windows for image acquisitions.²⁻⁵ While conventional SPECT has similar performance to PET, a major disadvantage of SPECT, among others, is the limited count sensitivity.^{2,5,6} This inherent limitation is in part attributed to the collimation techniques commonly used in SPECT imaging.

Multipinhole collimation has recently gained a lot of attention in the effort to develop high sensitivity and high resolution SPECT imaging systems.⁶⁻¹¹ In comparison to con-

ventional pinhole imaging, SPECT imaging with multipinhole collimators offers substantially higher count sensitivity while maintaining high image resolution. A trade-off exists, however, since multipinhole collimators provide a smaller field of view for the same detector size.

Coded aperture imaging is a technique originally proposed for astronomical imaging, where the incident rays from a point source are parallel to each other.¹² In medical applications, however, the objects are placed close to the mask and the incident rays from a point source are diverging rather than parallel, resulting in severe near-field imaging artifacts that are characterized by strong background and nonuniform intensities across the field of view.¹³⁻¹⁵ Numerous research groups have investigated methods to reduce these near-field artifacts, including the mask/antimask dual acquisition approach proposed by Accorsi *et al.*^{13,15} and the correction method for aperture collimation effect developed by our research team.¹⁴ With these methods, high resolution planar images can be obtained using a near-field coded aperture technique.

To a certain extent, coded aperture can be considered as a special type of multipinhole collimator. However, in the context of SPECT imaging, coded aperture differs from multipinhole collimators in a number of aspects. In coded aperture, the pinholes are arranged in a special pattern so that a matching pattern, the decoding mask, exists mathematically. This leads to a unique processing step in coded aperture imaging—decoding, where the decoded image is obtained by correlating the raw image with the decoding mask. Except

for those adopting ring geometry for the pinhole arrangement,^{6,8,9} multipinhole systems typically employ 7–16 pinholes.^{7,10,11} For coded aperture, however, hundreds of pinholes are usually machined in a coded aperture mask. For instance, a mask of 62×62 no-two-holes-touching (NTHT) modified uniformly redundant array (MURA) may have an open fraction of up to 12.5%,¹³ i.e., 480 open apertures. The large numbers of open apertures inevitably result in massive photon overlapping or signal multiplexing in data acquisitions, which has been viewed as a detrimental factor in multipinhole SPECT imaging^{7,10,11} and special pinhole placements may be needed to reduce multiplexing artifacts.¹¹ On the other hand, the massive photon overlapping can be viewed as an advantage in terms of high count sensitivity and efficient detector usage, even though it remains a major challenge in near-field coded aperture imaging to demultiplex these overlapped signals, particularly in the three-dimensional (3D) space.

While substantial progress has been made to reduce near-field coded aperture image artifacts, 3D image reconstruction methods for coded aperture images remained elusive partly due to the variable magnification factor coupled with the complex decoding or deconvolution process. Results of partial 3D reconstruction from a single coded aperture projection have been reported previously by others.^{16–19} However, these methods produced low quality images with poor resolution along the depth direction. Meikle *et al.* reported their studies on coded aperture SPECT for small animal imaging and provided useful perspectives on numerous system designs.^{20,21} However, the results reported therein were obtained mostly from computer simulations.

In this article, we extend our previous research^{14,22} and present a novel 3D image reconstruction approach to SPECT based on near-field coded aperture collimation and maximum likelihood estimation. The article is organized as follows. In Sec. II, we briefly review the progress we made in our previous study and describe in detail our 3D image reconstruction methods. The experimental phantom results are presented in Sec. III, and discussions on various aspects of the new coded aperture SPECT systems and image reconstruction methods are provided in Sec. IV. Finally, we conclude with discussions of our findings in Sec. V.

II. METHODS

II.A. Image reconstruction algorithm

A common approach to reconstructing tomographical image slices from a single coded aperture projection is coded aperture laminography, where the slice at depth z is computed by¹⁶

$$\hat{f}(x, y, z) = p(x, y) \otimes g_z(x, y), \quad (1)$$

where $\otimes$ is the correlation operator and $\hat{f}(x, y, z)$, $p(x, y)$, and $g_z(x, y)$ represent reconstructed image slice at depth z , the raw coded aperture projection, and the decoding mask at depth z , respectively. This approach suffers from the defocus artifacts. More specifically, the reconstructed in-focus slice is

corrupted by the obscurities from the activities in out-of-focus slices.^{14,16} A study by Accorsi²³ on the 3D point spread function (PSF) of coded aperture laminography suggested that the amount of depth information contained in a coded aperture image depends on the angular coverage of the projection, which is usually limited by the gantry size of the gamma camera used. In our previous study, we developed a maximum likelihood expectation maximization (MLEM) based method¹⁴ to reconstruct a 3D object from a single coded aperture projection, which is discussed in more detail later in Sec. II A. We noticed that, similar to laminography, the stack of images reconstructed in this way suffered from poor resolution along the depth (z) direction because the mask shadows from the different depths are highly correlated (differing only in magnification), and the information contained in the single coded aperture projection is insufficient to reconstruct a 3D object adequately. To fully reconstruct a 3D object, multiple coded aperture images acquired from different angles may be required in the 3D image reconstruction.

In this section, we introduce a novel 3D image reconstruction algorithm as an extension of our previous work. This new method starts with a partial 3D coded aperture image reconstruction using individual coded aperture projections. Secondary projections are generated by summing up the reconstructed image stacks along the depth (z) direction, simulating parallel-hole collimated SPECT projections. Subsequently, 3D SPECT images of the object are reconstructed slice by slice from the secondary projections using the ordered subset expectation and maximization (OSEM) algorithm.²⁴

Figure 1 illustrates the processing of a coded aperture projection. Multiangular coded aperture projections are acquired by rotating the camera head or the object in a circular orbit [Fig. 1(a)]. Each coded aperture projection [Fig. 1(b)] is corrected for radioactive decay and image uniformity, and is in turn deconvolved using the MLEM algorithm to create a stack of two-dimensional (2D) images [Fig. 1(c)]. The deconvolved images each with the out-of-focus obscurities removed are subsequently corrected for magnification and summed along the z direction to create a secondary projection image [Fig. 1(d)]. This summing process is conceptually identical to the process of parallel-hole collimated imaging while preserving the high sensitivity and high resolution properties of coded aperture imaging. This process is repeated for each of the projections acquired, creating secondary projections from multiple angles. Sinograms of cross-sectional slices can be formed by extracting the corresponding columns from all secondary projections, which are subsequently used to reconstruct the slices with the OSEM algorithm. We describe below in detail the steps involved in the proposed image reconstruction scheme for 3D coded aperture SPECT.

II.A.1. Partial reconstruction of 3D image stack from single projection

As mentioned, the work presented in this paper is an extension of our previous effort on near-field coded aperture

FIG. 1. Illustration of the process to generate a secondary projection from a raw coded aperture projection. (a) Diagram of coded aperture imaging system; (b) raw coded aperture projection, $p(x,y)$; (c) image stack reconstructed from (b), $f(x,y)$; (d) secondary projection, I . Step 1: Image acquisition. Step 2: MLEM reconstruction using Eq. (2). Step 3: Simulating parallel reprojection by summing the images shown in (c) and correcting for magnification variation.

imaging.¹⁴ Here we briefly reiterate our previous methodology for completeness of the 3D image reconstruction method proposed. At focus is the method to reconstruct a 3D image stack from a single coded aperture projection, a critical component of the novel 3D image reconstruction method introduced herein.

As mentioned earlier, the application of coded aperture technique in medical imaging has been plagued by the near-field artifacts of nonuniform intensity across the field of view (FOV) and the background noise outside the object. The non-uniformity artifact, i.e., angular count sensitivity variation across the FOV, is in part caused by the collimation effect of coded aperture masks with finite thickness. In our previous work,¹⁴ we developed a correction method to compensate for this collimation effect and greatly reduced the near-field artifact without the mask/antimask dual image acquisitions.¹⁵

A mosaic of four identical mask patterns is commonly used in conventional coded aperture imaging and a decoding mask is designed and correlated with the central quadrant of the acquired image to obtain a decoded image.^{15,25} In our previous work, we adopted a different approach and applied the MLEM method to deconvolve the entire coded aperture image acquired.¹⁴ We found that the MLEM method coupled with our aperture collimation correction had the advantages of imposing an inherent non-negativity constraint on image intensity, eliminating aliasing artifacts, largely reducing the background noise outside the object, and yielding much improved image reconstruction quality. In our previous phantom experiments, we obtained images with excellent in-plane (intrinsic spatial) resolution (<1 mm) for 2D thin objects.¹⁴ However, for a thick object the deconvolved image quality deteriorated because magnification varied at different depths in the thick object. To account for this depth dependent magnification, we developed a modified MLEM method to reconstruct a 3D image stack from a single coded aperture projection. In this method, the image slice at depth z is updated by^{14,22}

$$\hat{f}^{(k+1)}(z) = \frac{\hat{f}^{(k)}(z)}{\sum_z h(z)} \left\{ h(z) \otimes \frac{p - \sum_{z' \neq z} [\hat{f}^{(k)}(z') * h(z')]}{\hat{f}^{(k)}(z) * h(z)} \right\}, \quad (2)$$

where $\otimes$ denotes the correlation operator, $*$ denotes the convolution operator, z represents the object-to-detector dis-

tance, p represents the coded aperture projection after the collimation correction, $\hat{f}^{(k)}(z)$ is an estimate of the object at depth (slice) z after k iterations, and $h(z)$ is the coded aperture mask shadow corresponding to z . Note that the in-plane coordinates, (x,y) where x is parallel to the axis of rotation and y is perpendicular to the x (see Fig. 1), are omitted in Eq. (2) to simplify the expression. The process expressed in Eq. (2) differs from the conventional MLEM deconvolution in that the expected contribution from the “out-of-focus” slices ($z' \neq z$) is subtracted from the measured projection, and the correction ratio in the division step is calculated only for the “in-focus” slice. More specifically, the correction ratio is computed only from the estimation errors in the in-focus slice. Hence, the algorithm is expected to converge faster. To avoid negative pixels, negative elements in the updating factor calculated in the correlation step in Eq. (2) are replaced by a small positive value (e.g., 10^{-6}). We had shown in our previous phantom studies that this method was capable of partially restoring the depth information of a 3D object, resolving two planar objects separated by 4 cm along the z direction.¹⁴

II.A.2. Correction for variable magnification in slice summation

As indicated earlier, each of the secondary projections is created by summing the corresponding reconstructed image stack along the z direction. This summing procedure is not quite a straight element-by-element summation because the magnification factors are different for slices at different depths and the pixel sizes of the reconstructed slices in the 3D image stack are also different. This magnification effect is illustrated in Fig. 2, where a denotes the object-to-mask distance and b the mask-to-detector distance. Consider a point source at P casting a mask shadow onto the detector area between D_1 and D_2 as shown in Fig. 2. When the point source is shifted to another point at the same depth, P' , the mask shadow is shifted to the area between D_1' and D_2' . It can be shown that the distance between D_1 and D_1' is equal to that between D_2 and D_2' . Using the geometry of similar triangles, we yield

FIG. 2. Illustration of the depth dependent pixel size in reconstructed image slices. Point sources at two points of the same depth, P and P' , project the mask shadows at D_1D_2 and $D_1'D_2'$, respectively. When the distance between D_1 and D_1' equals the pixel size of the raw coded aperture image, the distance between P and P' represents the pixel size of the reconstructed image slice at that depth.

$$\overline{PP'} = \frac{a}{b} \overline{D_1D_1'} = \frac{z-b}{b} \overline{D_1D_1'}, \quad (3)$$

where $\overline{PP'}$ and $\overline{D_1D_1'}$, respectively, represent the distance between points P and P' and the distance between D_1 and D_1' . Based on the definition of convolution, the pixel size at the object plane is the distance between P and P' when the distance between D_1 and D_1' is equal to the pixel size of p . Therefore, if the acquired coded aperture projection has a pixel size of δp , then the pixel size of slice z in the reconstructed image stack, $\delta f(z)$, is given by

$$\delta f(z) = \left(\frac{z}{b} - 1 \right) \delta p. \quad (4)$$

Equation (4) shows that the pixel size increases as z increases. To account for the variable pixel size in the image stack, prior to the summation each slice is resampled by interpolation with a sampling space equal to the pixel size of a particular slice, which is usually selected to be the slice that is the closest to the mask in order to preserve in-plane (intrinsic spatial) resolution.

After the magnification correction, a secondary projection is calculated by summing the slices along the z axis as follows. Let $f_n^*(x, y, z)$ be the reconstructed 3D image at projection angle n after the magnification correction; the secondary projection, $I_n(x, y)$, is given by

$$I_n(x, y) = \sum_z f_n^*(x, y, z). \quad (5)$$

II.A.3. Sinogram generation

In the secondary projections, the columns of the sinogram are perpendicular to the axis of rotation (see Fig. 1), and each column represents the summed projection of a cross-sectional slice in the 3D object. A sinogram²⁶ of a cross-sectional slice at $x=X$, $S_X(n, y)$, can be formed by extracting the corresponding columns from all secondary projections and assembling them side by side,

FIG. 3. A diagram of digitized basic pattern of the mosaic coded aperture mask used in the experiments.

$$S_X(n, y) = I_n(X, y). \quad (6)$$

Consequently, each sinogram represents the projections of a cross-sectional slice of the object from all angles and each column of the sinogram represents the projection of that slice at a particular angle.

II.A.4. Full 3D image reconstruction via OSEM

Three-dimensional SPECT images are reconstructed slice by slice from the sinograms generated above. Based on the known imaging geometry, the projections are divided into a number of subsets and the image slices are reconstructed using the OSEM algorithm²⁴ originally developed for parallel-hole collimated projections.

II.B. Coded aperture imaging systems and image acquisition

A dual-head SPECT imaging system (Varicam, GE Medical Systems, Waukesha, WI) with an intrinsic resolution of 3.9 mm was used in our image acquisitions. The dual-head SPECT camera was equipped with a coded aperture module on one head and a pinhole collimator on the other head. For the pinhole collimator, the diameter of the pinhole was 1 mm and the pinhole-to-detector distance was 12 cm. For the coded aperture module, the mask was placed in parallel to the detector and at 32.2 cm from the detector. The basic pattern of the coded aperture mask was a 46×46 NTHT (Ref. 15) MURA.^{27,28} The aperture diameter was 1.1 mm and the mask thickness was 2.0 mm. A diagram of the basic pattern of the coded aperture mask digitized is shown in Fig. 3. The size of the basic mask pattern was $5.06 \times 5.06 \text{ cm}^2$ and the full mask consisted of a mosaic of four basic patterns. The radiation sensitive area on the detector heads was $39 \times 51 \text{ cm}^2$. An image matrix of 512×512 with a pixel size of $1.105 \times 1.105 \text{ mm}^2$ was used in image acquisitions.

Multiangular projections were acquired by rotating the object instead of the camera heads because the coded aperture mask was too close to the camera rotating axis. More specifically, the maximal camera arm extension was 68 cm

FIG. 4. Schematics of the experimental imaging system. A phantom is secured by a metal bar to the stepping motor which allows for a circular rotation to acquire multiangular projections. The rotating axis of the phantom is visually aligned with the axis of rotation of the camera heads.

and the mask-to-detector distance was 32.2 cm, leaving only a cylindrical shape space with a diameter less than 4 cm for the object. Although rotating the object in one direction is conceptually equivalent to rotating the camera heads in the opposite direction, in reality it is not a trivial task to establish this equivalence because the object rotating axis needs to be coincided with that of the camera heads. Some image degradation is expected in the reconstructed images due to the potential axis misalignment. Figure 4 depicts the imaging systems used in our phantom experiments. The object was secured by a metal bar attached to a stepping motor, and the stepping motor was placed on an L-shaped frame clamped on the imaging table. Before image acquisitions, the object's rotating axis was first checked using a level to ensure that the axis was parallel to the detector surface, and the axis was then visually aligned with the rows of the pinhole image seen on the image acquisition monitor.

III. Phantom experiments and results

III.A. Experimental setup and results for pyramid shaped capillary tube phantom

As the first attempt to test the coded aperture imaging system and evaluate the performance of our 3D image reconstruction methods, we constructed a simple 3D pyramid shaped phantom. The phantom was made of four capillary tubes, each of which had a length of 6.5 cm, an inner diameter of 1 mm, and a wall thickness of 0.2 mm. The capillary tubes were each filled with 75 μCi of $^{99\text{m}}\text{Tc}$ solution and attached radially along the slope to the inner wall of a small funnel, as demonstrated in Fig. 4. The metal bar attached to the rotating motor was screwed into the cylindrical opening of the funnel to secure the phantom. Because the conic part of the funnel had an opening angle of approximately 60° , the cross section of this object should include four ellipse-shaped point sources with different separations. This phantom was placed in such a way that its central axis (rotating axis) was 5.5 cm away from the pinhole and 18.5 cm away from the coded aperture mask. Pinhole and coded aperture

FIG. 5. Reconstructed image slices at object-to-mask distances from 165 to 205 mm. The image stack was reconstructed from a single raw coded aperture projection of the pyramid shaped capillary tube phantom.

projections were acquired simultaneously over a full 360° phantom rotation using a step-and-shoot imaging protocol (2 min/step). A total of 48 projections (24 from pinhole plus 24 from coded aperture) were acquired. As compared to pinhole collimated imaging, coded aperture imaging resulted in a 50-fold increase in count sensitivity in this phantom experiment (1.1×10^6 counts/projection vs 2.2×10^4 counts/projection), despite the much larger object-to-mask distance for the coded aperture module.

For each coded aperture projection, a 3D image stack was reconstructed using Eq. (2). A total of nine images in each image stack were reconstructed using 200 iterations of the MLEM algorithm. The large number of iterations is needed because the image stack was reconstructed from a single projection, and the algorithm was expected to converge at a lower speed than the OSEM using multiangular projections. The execution time of the prototype codes implemented with MATLAB (MathWorks, Inc., Natick, MA) on a PC with a Pentium IV 2.6 GHz CPU was approximately 1 s per iteration. Each of the reconstructed images in the stack corresponds to an object depth (5 mm slice thickness) ranging from 165 to 205 mm away from the mask. The slice thickness of 5 mm is selected empirically as a result of a trade-off between computational efficiency and the effectiveness of removing the out-of-focus blur. Figure 5 shows one of these reconstructed image stacks. In the next step, a secondary projection was generated from each of the image stacks after the magnification correction and image summation processes were performed. Shown in Fig. 6 are two representative secondary projections and the corresponding pinhole collimated images in orthogonal views. In particular, Fig. 6(a) was ob-

FIG. 6. Examples of [(a) and (b)] secondary projections and [(c) and (d)] the corresponding pinhole collimated images. The raw coded aperture projections used in the reconstructions of (a) and (b) were acquired simultaneously with the pinhole images shown in (c) and (d), respectively.

tained from the image stack shown in Fig. 5. Compared with the pinhole collimated projections, the secondary projections appear to be at a slightly lower resolution in part due to the fact that the object was placed at a distance farther away from the coded aperture mask. In the coded aperture image acquisition, the object-to-detector distance was 50.7 cm, resulting in a magnification factor of only 2.74, while in the pinhole collimated imaging the object-to-detector distance was 17.5 cm with a magnification factor of 3.18. Note that magnification factor in coded aperture imaging represents the magnification of the mask shadow,^{14,15} given by $m_C = z/a$ (see Fig. 2), which is different from that defined in the context of pinhole imaging, given by $m_P = b/a = m_C - 1$. In this paper, we adopt the coded aperture imaging notation in reporting the magnification factors for pinhole imaging as well. While the background outside of the object was mostly suppressed through coded aperture image reconstruction [see Figs. 6(a) and 6(b)], the benefit of pinhole collimation appears to be the suppressed background artifacts between the capillary tubes [see Figs. 6(c) and 6(d)]. Note that the secondary projections were emulated parallel-hole collimated projections. Therefore, the capillary tubes are at different lengths from those in the pinhole collimated projections because of the difference in projection geometry.

After the secondary projections were generated, the images were compensated for decay and the sinograms of the cross-sectional slices were formed as mentioned in Sec. II A. Two representative sinograms corresponding to slices 60 and 90 of the secondary projections are shown in Fig. 7. The sinograms consisted of 24 angular projections each with 128

FIG. 7. Representative sinograms of two cross-sectional slices.

samples (pixels). Note that the original image matrix size of the projections (512×512) was in fact larger than 128 pixels. We extracted only 128 pixels from the central portion consisting of the objects to reduce the computation time. In this step, the 128 pixels had to be selected carefully to ensure the equivalence of object rotation to camera rotation, so that the sinograms were vertically symmetric, matching the desired rotating geometry.

Cross-sectional images of the capillary tube phantom were reconstructed using the OSEM with four subsets and five iterations. Figure 8 shows six representative slices from the reconstructed 3D images. As expected, the slice images show clearly four objects with decreasing separations. In the reconstructed images, the voxel size was calculated by Eq. (4) as 0.57 mm based on the in-plane pixel size in the nearest slice to the mask (165 mm). As seen in Fig. 8, the reconstructed images are of excellent image resolution, exhibiting much reduced background noise between the objects as compared to the secondary projections shown in Fig. 6.

III.B. Experimental setup and results for micro hot rod phantom

To further evaluate the performance of our coded aperture imaging system and 3D SPECT reconstruction methods, a micro hot rod phantom (Data Spectrum, Hillsborough, NC) was used in our experiment. The phantom had an inner diameter of 4.5 cm and a height of 6.3 cm and consisted of one large central rod and six groups of microrods arranged in a triangular shape as shown in Fig. 9(a). The rods in the six groups had diameters of 1.2, 1.6, 2.4, 3.2, 4.0, and 4.8 mm, and the center-to-center spacings within each group were twice of the rod diameter in the corresponding group. There were 3 rods in the 4.0 and 4.8 mm groups, 6 rods in the 3.2 mm group, 10 rods in the 2.4 mm group, 19 rods in the 1.6 mm group, and 34 rods in the 1.2 mm group.

FIG. 8. Cross-sectional images reconstructed from 24 angular projections of the capillary tube phantom using the OSEM (four subsets and five iterations). The separation between the capillary tubes decreases from slice 50 to slice 100, reflecting the pyramid shape of the phantom. Note that the background artifact between the capillary tubes, seen in the secondary projections (Fig. 6), are largely suppressed after the 3D image reconstruction.

As similar to the experiment described in Sec. III A, the micro hot rod phantom filled with approximately 900 μCi of $^{99\text{m}}\text{Tc}$ solution was attached to the metal bar so that the object could be rotated around its central axis by the stepping motor. The distance between the central axis of the phantom and the coded aperture mask was 18 cm. A total of 60 projections over a full 360° phantom rotation were acquired using a step-and-shoot imaging protocol (2 min/step). A total of nine slices each with a 5 mm thickness, located at 16–20 cm from the mask, were reconstructed using the same methods described in the previous section with the deviations described below. The secondary projections were rebinned by a factor of 2 to form 256×256 matrices. As a result, the pixel size became $1.04 \times 1.04 \text{ mm}^2$ and the number of samples for the sinograms was reduced by half, from

128 to 64. Because the hot rods insert [Fig. 9(a)] was composed of small cylindrical shaped rods, the cross sections of the phantom were identical. In the OSEM reconstruction for the cross-sectional image of this phantom, we used the summed sinogram from 20 consecutive slices to improve the signal-to-noise ratio (SNR). Figure 9(b) shows the image reconstructed from the summed sinogram using the OSEM with ten subsets and ten iterations. As seen, the six groups of rods in triangular shape can be identified visually from the reconstructed image. In addition, the hot rods in four of the groups, with rod diameters ranging from 2.4 to 4.8 mm, can be visually resolved. This result is particularly encouraging given the suboptimal imaging conditions including the small magnification due to the large object-to-mask distance and the possible errors introduced by the imperfect rotating mechanism.

FIG. 9. Experimental results of micro hot rod phantom. A total of 60 evenly distributed angular projections were acquired over a full 360° phantom rotation, each with a 2 min acquisition time. (a) Picture of the phantom; (b) cross-sectional image reconstructed by the OSEM (ten subsets and ten iterations) from the summed sinogram of 20 consecutive cross-sectional slices.

IV. DISCUSSIONS

We have demonstrated with experimental results the feasibilities of our near-field coded aperture imaging technique and novel image reconstruction methods for 3D SPECT. While the reconstructed image quality obtained from the preliminary phantom results is quite promising, a number of limitations in our experimental systems have undermined our efforts to improve the image reconstruction. Among them, the most prominent limitation was the large mask-to-detector distance of the coded aperture module, which forced us to rotate the object instead of the cameras during image acquisitions. Detailed system calibrations including precise rotation axis alignment were not performed in this study. As a result, the geometry of cross-sectional slices of the object varied from one projection to another and the imperfect im-

aging geometry inevitably corrupted the quality of image reconstruction, particularly for a complex object such as the micro hot rod phantom used in this study.

In our phantom experiments, the objects had to be placed quite far from the detector to avoid image truncation¹⁴ due to the large size of the mask used, resulting in a small magnification. Additional sources of error were the imperfect object placement and rotating mechanism as mentioned above, which in turn introduced degradation in the image reconstruction. As such, given the early stage of the research and the imperfection of the experimental system, it would be premature to perform quantitative analysis on image reconstruction quality or comparison with other existing imaging modalities, such as pinhole or multipinhole collimated 3D SPECT imaging. However, we feel that the results presented herein are quite encouraging, particularly in the situations of the existing system limitations and the small number of projections used in our 3D SPECT reconstruction.

Despite the limitations above, one should not perceive these limitations as shortcomings of coded aperture SPECT imaging because image reconstruction of superior quality can be achieved with a better designed mask. For example, in the phantom experiment of four capillary tubes, the magnification factor at the center slice was only 2.74. In the future mask design, we may choose to reduce the mask-to-detector distance from the current 32.2 to 24 cm and use a single basic mask pattern, i.e., the mask size of $5.06 \times 5.06 \text{ cm}^2$, and a thinner mask, e.g., 1.0 mm. With a rotation radius (i.e., object-mask distance) of 8 cm, the new system design will allow for a greater magnification factor of 4 (i.e., $24/8+1$) without image truncation. Although the redesigned mask size will be reduced to half of the mask size that we used in this study, the sensitivity may in fact increase because the object can be placed much closer to the detector. System design is one of our future research directions and a thorough discussion on this topic is beyond the scope of this paper. Nonetheless, we expect that an improved design of coded aperture imaging systems will likely improve the quality of image reconstruction.

The experiment presented in Sec. III A had shown that the coded aperture mask yielded a count sensitivity orders of magnitude higher than that of a single-pinhole collimator. In general, higher count sensitivity leads to better SNR in the images because of the Poisson statistics of photon counting processes. However, when comparing count sensitivity between different imaging modalities, one also needs to take into account the image reconstruction process. In our case, the image reconstruction for coded aperture imaging is substantially more complicated than that of the single- or multipinhole systems. In 2D image reconstruction, because the coded aperture mask is well conditioned, the noise amplification in the reconstruction process is well controlled, usually resulting in improved SNR in the reconstructed images. Noise propagation in 3D image reconstruction, however, is much more complex and deserves further investigation. While coded aperture technique offers substantially higher count sensitivity, it is important to note that count sensitivity increase does not always translate into improvement on SNR

of the same magnitude in the reconstructed images, and its benefit on image reconstruction quality needs to be evaluated.

The primary goal of this study was to demonstrate the feasibility of 3D SPECT imaging with the near-field coded aperture imaging technique. The 3D image reconstruction algorithm introduced herein employed the MLEM-based partial 3D image reconstruction algorithm using a single coded aperture projection that we previously developed.¹⁴ We further adapted the OSEM algorithm²⁴ originally designed for parallel-hole collimated SPECT to provide a fast path to full 3D coded aperture image reconstruction. Aside from the implementation convenience, the proposed two-step approach has several advantages compared to the direct MLEM reconstruction method widely used for multipinhole SPECT. Direct MLEM method requires detailed knowledge of the transition matrix through either measurements or mathematical modeling. The coded aperture mask used in our experiments has hundreds of apertures, considerably more complicated than multipinhole collimators. Accurately modeling or measuring the transition matrix of such a complex mask at various projection angles is a nontrivial task. More importantly, the transition matrix is a $K \times M$ matrix where K is the number of voxels in the target volume and M is the number of measurement positions. For a target volume of $128 \times 128 \times 128$ voxels, $K=2^{21}$ ($\sim 2 \times 10^6$). In our first experiment in which 24 projections each with a matrix size of 512×512 were used in the 3D image reconstruction, $M=24 \times 512 \times 512$ ($\sim 6 \times 10^6$). It would be prohibitive to store a matrix of such size (12×10^{12}) even with a state-of-the-art computing system. Multipinhole systems may not suffer from this restriction because their transition matrices are sparse, i.e., having only a small fraction of nonzero elements. However, this is not the case for the transition matrix of a coded aperture system, which has significantly more nonzero elements due to massive overlapping. Adding to the dilemma is the lack of flexibility, i.e., the transition matrix is specific to the imaging protocol. For example, the transition matrix for the experiment in Sec. III A would not be applicable for the experiment in Sec. III B because the projection angles were different. While this obstacle may be overcome by calculating the relevant elements as they are needed in the OSEM reconstruction, this approach would be inefficient and impractical for coded aperture because the masks are substantially more complex than multipinhole collimators and it would require longer time to calculate those elements on the fly. In contrast, the proposed approach decouples the 3D reconstruction problem into two steps, resulting in significant data reduction as well as computing efficiency. The partial 3D reconstruction in Eq. (2) uses only the mask shadow, which is derived from the mask matrix and the magnification factor. This step is identical for all projections in which the same mask shadows are used, providing the desired flexibility. The next step, i.e., the OSEM step, is well established and the implementation is straightforward. Above all, the proposed method does not require measurement or storage of a terabyte sized transition matrix.

It is also worthwhile to note that the investigation on the 3D reconstruction algorithm remains preliminary in this study. More comprehensive evaluations are needed. For example, an experiment with a cold rod phantom would be of interest, in particular, to cardiac applications in which imaging of myocardial perfusion defects is a general clinical objective. The SNR and contrast recovery in the reconstructed coded aperture images were not evaluated in this study, and the performance of coded aperture imaging in comparison with common (multi) pinhole imaging has yet to be investigated. This comparison is beyond the scope of this paper but will be investigated in the future.

The phantom results demonstrated in Sec. III were obtained in “one pass” of the image reconstruction procedure, i.e., from the MLEM reconstruction of 3D image stacks to the OSEM reconstruction of cross-sectional slices. We anticipate that further improvements may be achieved by iteratively feeding the 3D SPECT volume reconstructed slice by slice from the sinograms back to the MLEM reconstruction. More specifically, for each acquired projection, the OSEM reconstructed SPECT volume can be rotated and resampled to create an image stack which can be in turn used as the initial guess, $f^{(0)}$, in the MLEM reconstruction process formulated in Eq. (2). As shown in Sec. III A, background activities remain noticeable between the capillary tubes in the secondary projections (Fig. 6), whereas they are largely reduced in the OSEM reconstructed slices (Fig. 8), thanks to the image reconstruction from multiangular projections. By feeding back the reconstructed 3D SPECT volume, a substantial increase in signal-to-background ratio in the secondary projections can be expected and thus better quality of 3D SPECT reconstruction can be obtained. The trade-off may be the overhead involved in rotating and resampling the 3D volume, which needs to be justified through further experimentation.

Another direction of our future research in 3D coded aperture SPECT imaging is to reduce the number of projections for image reconstruction and to optimize the systems for small animal imaging. Our current coded aperture mask was originally designed for 2D imaging studies, where large mask-to-detector and mask-to-object distances were employed to minimize the blurring artifact caused by the object thickness. As a result, the projection angle spanned by the mask at such a distance is rather small, i.e., the differences among the view angles from the apertures on the mask are small. Consequently, projections from a large number of angles are required for high quality 3D image reconstruction, demanding a long image acquisition time that may limit the application of this imaging technology. For small animal imaging where only a small field of view is needed, the mask can be moved closer to the detector and the object can be placed closer to the mask so that a wider projection angle spanned by the mask can be achieved. By doing so, we can explore the possibility of reconstructing a 3D object with a small number of projections.

V. CONCLUSION

We have developed a novel 3D image reconstruction approach to near-field coded aperture SPECT and evaluated our methods using a customized capillary tube phantom as well as a complex micro hot rod phantom. The experimental results have demonstrated empirically that our approaches to near-field coded aperture imaging and MLEM image reconstruction may have the potential for high sensitivity and high resolution SPECT. To our knowledge, the methods and phantom study presented herein were the first SPECT reconstruction methods developed for multiangular near-field coded aperture imaging and also the first study using real phantom data in 3D coded aperture SPECT reconstruction. Future studies on the optimization of coded aperture mask design, algorithms for the 3D reconstruction of coded aperture SPECT with a small number of projections, and the correction for photon attenuation are warranted for the systems to be readily used in clinical applications.

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