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Spatial resolution properties in cone beam CT: A simulation study

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Abstract

This work is intended to investigate the spatial resolution properties in cone beam CT by estimating the point spread functions (PSFs) in the reconstructed 3D images through simulation. The point objects were modeled as 3D delta functions. Their projections onto the detector plane were analytically derived and blurred with 2D PSFs estimated and used to represent the detector and focal spot blurring effects. The 2D PSF for detector blurring was computed from the line spread function measured for a typical *a*-Si/CsI flat panel detector used for general radiography. The focal spot blurring effect was simulated for an x-ray source with a nominal focal spot size of 0.6 mm and 1.33 $\times$ magnification at the rotating center. Projection images were computed and sampled with an interval significantly smaller than the detector pixel size to avoid aliasing. Images were reconstructed using the Feldkamp algorithm with the five different filter functions. Reconstructed PSFs were plotted and analyzed to investigate the effects of detector blurring alone, focal spot blurring alone, or a combination of the two on the PSFs and their variations with the radial distance and *z*-level. Effects of binning and reconstruction filters were also studied. Our results show that the PSFs due to detector blurring are largely symmetric and vary little with the locations of the point objects. With focal spot blurring only or added to detector blurring, the PSFs along the rotation axis were largely symmetric but became increasingly asymmetric as the point objects were moved away from the rotation axis. The PSFs were found to become wider in the axial (anode to cathode) direction as the objects were moved toward the cathode side. The 3D PSFs may be approximated by an ellipsoid with three different axial lengths. They were found to point upright along the rotating axis but tilt toward the rotating axis as the point object was moved away from the axis.

Keywords

cone beam CT; spatial resolution; point spread function; detector blurring; focal spot blurring; Feldkamp reconstruction algorithm

I. INTRODUCTION

Conventional CT has relied on the use of collimated x-ray beam and multiple detector arrays in conjunction with table translation to obtain 3D images of the patient. Due to the large pixel size of the detector arrays used and the use of table translation, the spatial resolution of conventional CT images is generally limited and asymmetric, with the resolution along the axial or scanning direction often significantly lower than those in the transverse directions.

^{1,2} As a result, 2D point spread functions (PSFs) are often measured to quantify the spatial resolution in the transverse directions while the scan thickness or the x-ray intensity profile in the axial direction are often measured to characterize the axial resolution.

Following the commercialization of the flat panel detectors, there has been interest and effort in developing cone beam CT techniques for applications difficult for the conventional CT techniques to fit in. Examples include dedicated breast CT³⁻⁶ and CT integrated with surgery⁷ or radiation treatment devices for beam verification and treatment planning.⁸⁻¹⁰ One advantage of flat panel based cone beam CT is the nearly isotropic spatial resolution. Thus, the spatial resolution of a cone beam CT system needs to be characterized and measured in a 3D manner.

Spatial resolution in cone beam CT has been previously investigated and reported on.¹¹⁻¹⁵ Yan *et al.* have derived an expression for the PSF and MTF for cone beam CT. However, the derivation did not take into account the focal spot blurring effect. Chen *et al.*¹⁶ reported on the use of images of metal beads to estimate the PSFs by modeling them as a spherically symmetric Gaussian function and using an iterative edge-blurring algorithm. Their method seems to apply to spherically symmetric PSFs only and they demonstrated the method by measuring the width of the PSF at three different locations along the rotating axis. Kwan *et al.*¹³ have recently reported on the evaluation of the spatial resolution characteristics of a cone beam breast scanner. They used slightly tilted thin metal wires to measure the MTFs in the coronal plan and along the axial axis and investigated the effects of the radial distance, cone angle, voxel size, and reconstruction filters. Their study was mainly intended for an experimental cone beam breast CT scanner that employs a small focal spot tube, a *a*-Si/CsI flat panel detector operated in 2×2 binning mode, and 1.92× magnification at the rotating center. MTF differences between the coronal plane and the axial direction were demonstrated, but the MTFs along the axial direction and their dependence on the cone angle were not investigated.

It is well known that in the absence of the patient scatter, two processes contribute to the degradation of the spatial resolution in acquiring the projection images: detector blurring and focal spot blurring. These two processes are very different in nature. The detector blurring effect does not vary with location and may be represented by a 2D PSF. The focal spot blurring effect, on the other hand, varies with the location in both the axial (anode to cathode) as well as transverse directions. This results in a 2D PSF whose width in the axial direction increases toward the cathode and decreases toward the anode, resulting in position dependent and asymmetric PSFs. Furthermore, the PSFs are upright only along the projected rotating axis. The off-axis PSFs become tilted away from the axis with the degree varying with the distance from the axis. Since the detector, focal spot size, and imaging geometry are usually selected in such a way that none of the two blurring effects dominates the other, these two different types of blurring effects together determine the PSFs in the 3D reconstructed images. Thus, it is interesting and important to understand the effects of the two blurring processes alone by themselves and those of the blurring processes together. Such a study would be difficult and tedious with imaging experiments, as it is impossible to isolate the two blurring effects from each other. A ray tracing simulation method, on the other hand, would allow the two blurring effects to be separately estimated for comparison or combined together to simulate the real imaging situations.

In this study, we used a ray tracing based simulation method to investigate and estimate the 3D PSFs in the reconstructed cone beam CT images. Projection images of point objects were simulated to investigate the effects of detector blurring alone, focal spot blurring alone, and detector and focal spot blurring together on the PSFs. The variation of the PSFs with the radial distance and the cone angle were also estimated and investigated.

II. THEORY AND METHODS

The PSF in the reconstructed cone beam CT images may be affected by a number of factors associated with the blurring of the point objects in forming the projection images and those associated with the reconstruction process. In our simulation study, point objects at selected locations were projected onto the detector along with the focal spot blurring effect (Fig. 1). Detector blurring was modeled and added to compute the overall blurred projection images for various views. The results were then used to reconstruct and use the 3D images of the point objects to compute the 3D PSFs. In the following sections, the methods for the simulation study are described and discussed separately for each individual step.

II.A. Projection of point like objects

In order to estimate the PSFs for the reconstructed cone beam CT images, the blurring of a point-like object needs to be modeled and incorporated with the simulated projection images (Fig. 1). A point object, M , with the coordinates (x_0, y_0, z_0) , may be modeled as a delta function in the 3D space. The projection of M onto the detector plane is illustrated in Fig. 2 where the x-ray source-detector gantry is assumed to rotate around the origin, O , in the x - y plane but aligned with the x -axis in the illustrated geometry with the focal spot at $(d_{OS}, 0, 0)$ and the detector plane intersecting with the x -axis at $(-d_{OD}, 0, 0)$. Let the projection of M onto the detector plane be expressed as a two dimensional delta function as follows:¹⁷

$$P(u, v) = k \cdot \delta\left(u - \frac{d_{SD}}{d_{SD} - x_0} \cdot y_0, v - \frac{d_{SD}}{d_{SD} - x_0} \cdot z_0\right), \quad (1)$$

where u, v are the coordinates in the detector plane with the u and v axes parallel to the y, z axes and the origin located axis at $(-d_{OD}, 0, 0)$. k is a proportionality constant; (u_0, v_0) is the point at which M is projected to in the detector plane. Its coordinates may be expressed as follows:

$$u_0 = \frac{d_{SD}}{d_{SD} - x_0} \cdot y_0, v_0 = \frac{d_{SD}}{d_{SD} - x_0} \cdot z_0.$$

II.B. Detector blurring

Modern cone beam CT systems for patient imaging usually employ a flat panel detector for image acquisition to simplify computations for backprojection. There are two types of flat panel detectors: the direct detectors convert x rays directly into charge signals for readout and the indirect detectors employ a scintillator to convert the x rays into light photons, which are then converted into charge signals in the photodiode of each detector element. The spatial resolution of an indirect detector depends on both the blurring characteristics of the scintillator and the signal averaging aperture (characterized approximately by the pixel size), while that of a direct detector depends mainly on the latter. In this article, we assume that an indirect detector consisting of a $600 \mu\text{m}$ thickness thallium doped cesium iodide scintillation layer coupled to an active matrix photodiode array is used.¹⁸ The spatial resolution of an imaging system or component is often characterized by measuring the MTF in the frequency domain or the PSF in the spatial domain. The overall MTF of a cascaded imaging system is the product of the MTFs for individual stages. In the spatial domain, the overall PSF is the convolution of the PSFs for individual stages. Thus, the overall PSF of an indirect detector, $G_d(u, v)$, may be expressed as

$$G_d(u, v) = G_s(u, v) \otimes R_p(u, v), \quad (2)$$

where $\otimes$ is a convolution operator; $G_s(u, v)$ and $R_p(u, v)$ are the PSFs for the scintillation layer and signal averaging in the pixel area, respectively.

In the scintillation layer, the light photons converted from the x-ray photons are scattered in all directions and the resulting light intensity signals spread out away from the incident point. The PSF for this signal spread may be approximated by a 2D Gaussian function as follows:

$$G_s(u, v) = \frac{1}{2\pi\sigma_s^2} \exp\left(-\frac{u^2 + v^2}{2\sigma_s^2}\right), \quad (3)$$

where σ_s^2 is the variance of the Gaussian function.

Following the scintillating process, some of the light photons reach the photosensitive area in the flat panel—the photodiodes, and are converted into charge signals for later readout. The photodiode occupies a large fraction of the pixel area. The ratio of the photosensitive area to the pixel area is known as the fill factor. For detectors with a large pixel size, like those often used for cone beam CT, the fill factor is generally high (e.g., ~80%),¹⁸ with a linear dimension only slightly smaller (by ~10%) than the pixel size. For simplification, we assume that the image signals are smoothed over the entire pixel area and the corresponding blurring may be represented by a square aperture function whose dimensions are characterized by the pixel size. In the $m \times n$ pixel binning mode, the charge signals from $m \times n$ neighboring pixels are integrated together into a single signal. This is equivalent to the smoothing of the image signals with a rectangular aperture function with its dimensions enlarged by a factor of m and n in the horizontal and vertical directions, respectively. Thus, the aperture function for a detector operated in the $m \times n$ binning mode may be expressed as

$$R_p(u, v) = \frac{1}{mna^2} \cdot \text{rect}(u) \cdot \text{rect}(v) = \begin{cases} \frac{1}{mna^2} & |u| < \frac{ma}{2}, |v| < \frac{na}{2} \\ 0 & \text{elsewhere} \end{cases}, \quad (4)$$

where a is the pixel size; m and n are the numbers of binned pixels in the horizontal and vertical directions, respectively. $m=n=1$ when there is no binning.

Thus, the overall blurring function, $G_d(u, v)$, may be computed using Eqs. (2)-(4) if the parameters σ_s, m, n , and a , are known. To determine the σ_s in Eq. (3), m and n were set to 1 and a to 0.2 mm in Eq. (4). $G_d(u, v)$ was then fitted to the line spread function (LSF) measured for a radiographic detector with an unbinned ($m=n=1$) pixel size of 0.2 mm,¹⁹ resulting in a fitted value of 0.061 mm for σ_s . This value was used to compute $G_d(u, v)$ for various binning modes, which were then used to compute projections of the point objects for various view angles as follows:

$$P(u, v) = k \cdot \delta(u - u_0, v - v_0) \otimes G_s(u, v) \otimes R_p(u, v). \quad (5)$$

II.C. Focal spot blurring

It is well known that the dimension of the apparent focal spot in the cathode to anode direction becomes larger toward the cathode side and smaller toward the anode side. This

variation is insignificant in single slice and even multiple slice fan beam CT as the angular range for image coverage is generally narrow enough to ignore the variation. However, with the wide beam angle in cone beam CT, the apparent focal spot size varies significantly from the anode side to the cathode side in the projection images. The resulting variation of the focal spot blurring effect may affect the spatial resolution of the reconstructed images as well. One major goal of this work is therefore to investigate how this variation would affect the PSFs in the reconstructed cone beam CT images.

The simulation for the focal spot blurring effect for imaging a 3D object is a complex task. However, the simulation for the blurring of a point object is straightforward and requires only the intensity profile of the actual focal spot to be projected through the object point onto the detector plane, as illustrated in Fig. 3. To simplify the simulation task, we assume that the intensity profile for the actual focal spot on the target surface, tilted away from the anode to cathode direction by 10° , has the shape of a 2D Gaussian function

$$I_{fs}(y'', z'') = I_0 \exp\left(-\frac{y''^2}{2\sigma_{y''}^2} - \frac{z''^2}{2\sigma_{z''}^2}\right), \quad (6)$$

where I_0 is the maximum intensity; y'' and z'' are the coordinates in the tilted (by 10°) target surface with the origin at the center of the focal spot; and $\sigma_{y''}^2$ and $\sigma_{z''}^2$ are the variances of the Gaussian function along the y'' and z'' axes, respectively. With this model, the equal intensity contours would be elliptical. To simulate the x-ray source used in our experimental cone beam CT system with a nominal focal spot size of 0.6 mm, $\sigma_{y''}$ and $\sigma_{z''}$ were assumed to be 0.26 and 1.47 mm, corresponding to a full width at half maximum (FWHM) of 0.60 and 3.46 mm, respectively. The former is the nominal focal spot size of the x-ray source in our experimental system. This results in the nominal focal spot size of 0.61 mm in both the anode to cathode direction and the transverse direction at the center of the detector, which is assumed to point to the focal spot in a perpendicular-to-anode-to-cathode direction.

Using this focal spot to image a point object, the resultant image is essentially the projection of the focal spot through the object point onto the detector plane. To determine this projection, the center of the actual focal spot was first projected onto the detector plane. A rectangular region of interest (ROI) was then empirically defined to approximately encompass an area in which the projected intensity is above 1% of the maximum intensity. Each pixel inside this ROI was backprojected through the object point onto the tilted target surface. The intensity at the backprojection point was then corrected for the inverse square law using that at D as the reference and then assigned as the projected intensity for the specific pixel.

Similar effects of the inverse square law were also incorporated into the reference image with which the logarithmic attenuation data were computed and used for reconstruction. Notice that the effects of the inverse square law were largely canceled in computing the logarithmic attenuation data. However, slight intensity variations resulting from the differences in the distance of travel from various points in the focal spot to the detector would have remained and were incorporated in the computed projection data. For combined detector and focal spot blurring effects, the Gaussian function representing the detector blurring process was used to convolve the focal spot blurred image to further incorporate the detector blurring effect with the projection images. Projection images were computed and sampled with an interval ($15 \mu\text{m}$) significantly smaller than a typical detector pixel size ($194\text{--}388 \mu\text{m}$) used in CBCT to avoid aliasing.

II.D. FDK reconstruction

The Feldkamp algorithm was used for cone beam reconstruction in this study.²⁰ The algorithm is mainly based on the filtered backprojection method. Prior to filtering, the projection data were first weighted with a geometrical factor to form the weighted projection functions as follows:

$$P_{\beta}^{\text{weighted}}(u, v) = P_{\beta}(u, v) \frac{d_{so}}{\sqrt{d_{so}^2 + u^2 + v^2}}. \quad (6')$$

Then 1D convolution with a selected filter was performed row by row:

$$P_{\beta}^{\text{filtered}}(u, v) = P_{\beta}^{\text{weighted}}(u, v) \otimes h(u). \quad (7)$$

The reconstructed 3D images of the point objects were obtained by 3D backprojection of the filtered projections over all rotation angles:

$$PSF(x, y, z) = \int_0^{2\pi} \frac{d_{so}^2}{(d_{so} - y)^2} P_{\beta}^{\text{filtered}}\left(\frac{d_{so}}{d_{so} - y}x, \frac{d_{so}}{d_{so} - y}z\right) d\beta. \quad (8)$$

The 3D PSFs were computed from the reconstructed images of the point objects after normalizing the 3D signal profile for each object to 1.

Three hundred projection images were simulated every 1.2° interval and used for reconstruction. Various larger numbers of views were experimented with and found to have no effects on the resulting estimated PSFs. Following the experimentation, we chose to use 300 views to simplify the computation tasks without compromising the spatial resolution of the simulated reconstructed images. The projection data were reconstructed utilizing five different filters: Ramp (standard), Cosine, Hamming, Hanning, and Shepp–Logan filters. Each filter was defined according to the window that is needed to modify the ideal Ramp filter in frequency domain:

$$H(\omega) = |\omega|W(\omega), \quad (9)$$

where ω is the spatial frequency and $W(\omega)$ is defined as follows:

for Ramp filter:	$W(\omega)=1,$
for Cosine filter:	$W(\omega)=\cos(\omega/2),$
for Hamming filter:	$W(\omega)=0.54+0.46 \cos(\omega),$
for Hanning filter:	$W(\omega)=0.5+0.5 \cos(\omega),$
for Shepp–Logan filter:	$W(\omega)=\text{Sinc}(\omega/2).$

II.E. Comparison to experimental data

To validate the use of our simulation method for estimating the PSF, two 0.8 mm diam aluminum beads inserted in a cylindrical styrofoam block were imaged with cone beam CT.

The resulting 3D image data were compared to those computed by convolving an analytical model for the bead with the PSF determined by image simulation. A bench top experimental cone beam CT system was used for the imaging experiment. This system consists of a radiographic x-ray tube (Varian Medical Systems) with 0.6 mm focal spot size, *a*-Si/CsI flat panel detector (Paxscan 4030CB, Varian Medical Systems) and a step-motor-driven rotating table for holding and rotating the phantom. The x-ray source to detector and source to center distances were 100 cm and 75 cm, respectively, resulting in a magnification factor of 1.33 at the center of rotation. Three hundred projection images with an angular increment of 1.2° were acquired at 80 kVp and 32 mA with a pixel size of $194\ \mu\text{m}$. The Feldkamp algorithm was used in conjunction with a Ramp filter for image reconstruction with the voxel size set at $146\ \mu\text{m}$ (corresponding to $194\ \mu\text{m}$ pixel size backprojected to the center of rotation). The simulated 3D images were computed by analytically modeling the bead as a sphere of a high attenuation coefficient and then convolved with the PSFs determined by simulation with the same scanning and reconstruction parameters used. The simulated 3D image data were then compared to those reconstructed from actually acquired image data.

II.F. Display of 3D PSFs

Display of 3D PSFs in a 2D space is not straightforward. As will be clear from the results, the shape of the PSF varies with the location from the spherical shape to that close to a tilted, elongated ellipsoid. In this article, we chose to show the 3D PSFs by plotting the three orthogonal cross sections through the object points as grayscale images. These three cross sections were selected to be parallel to the *x*-*y*, *y*-*z*, and *x*-*z* planes and referred to as the x' - y' , y' - z' , and x' - z' planes. The corresponding cross-sectional images are referred to as the x' - y' , y' - z' , and x' - z' images. Similarly, the three axes in parallel to the *x*, *y*, and *z* axes but through the object point are referred to as x' , y' , z' axes, respectively. Notice that due to the rotating symmetry in CBCT using a 360° circular orbit (with *z*-axis as the rotating axis) for the focal spot, there is generally no difference in variations of the PSFs or plots along *x*, *y*, or any radial directions. For simplicity and consistency, we have chosen the radial direction to be along the *x*-axis and used it to demonstrate the variation of the PSFs and to plot the PSF values along the radial direction.

The 3D PSFs should be normalized in such a way that the integration of the PSFs is 1. However, to help illustrate the differences in sharpness and shape, the PSFs were normalized to 1 at the center of the PSFs, which was defined as where the point object was projected to in the projection image. Since the PSF values at the center are equal to or very close to their maxima, this helps bring the PSF values to the same range when plotting them as grayscale images or graphs to compare the PSFs for sharpness and shape.

III. RESULTS

Using the simulation methods described in Sec. II, we have computed the estimated PSFs for three different imaging situations: detector limited, focal spot limited, and detector/focal spot limited. In the detector limited situation, the focal spot blurring effect is substantially smaller than the detector blurring effect and can therefore be ignored. In the focal spot blurring limited situation, the detector blurring effect is substantially smaller than the focal spot blurring effect and can therefore be ignored. In the detector/focal spot limited situation, the detector blurring and focal spot blurring effects are of comparable magnitude and therefore need to be considered together.

III.A. Effects of detector blurring

The effects of detector blurring on the PSFs are shown by the grayscale images of two point objects for the x' - y' and y' - z' cross sections in Fig. 4: (a) x' - y' image of a point object at (0,

0, 0); (b) x' - y' image of a point object at (12, 0, 5) cm; (c) y' - z' image of a point object at (0, 0, 0); and (d) y' - z' image of a point object at (12, 0, 5) cm. The blurred object appeared to be symmetric and similar in size in all images. For quantitative comparison, the center [at (0, 0, 0)] and off-center [at (12, 0, 5) cm] PSFs were extracted from the data and analyzed. It was found that the difference between the center and off-center PSF values were largely insignificant along all directions. The FWHMs were computed from these profiles and found to be 0.13 and 0.12 mm along all directions for (0, 0, 0) and (12, 0, 5), respectively. Thus, the effect of detector blurring on the PSFs is largely a symmetric one, with the FWHMs a little smaller than the pixel size projected back to the center of rotation ($194 \mu\text{m}/1.33=146 \mu\text{m}$) in both axial and transverse directions. However, the off-center PSFs were found to be slightly sharper than the center PSFs along all directions.

III.B. Effects of binning

The effects of detector blurring and 2×4 binning are shown by the x' - y' and y' - z' images of a point object located at the rotation center in Figs. 5(a) and 5(b). The circular symmetry of the x' - y' image reflects the fact that the x' - y' plane was the rotation plane in which projection images of various view angles were acquired and superimposed through the backprojection process in reconstruction. The y' - z' image has a rectangular shape with the longer dimension in the axial or z -direction, reflecting the fact that image signals were integrated over a rectangular aperture (0.388 mm wide, 0.776 mm high) due to 2×4 binning.

III.C. Effects of focal spot blurring

Focal spot blurring degrades the spatial resolution of the projection images by blurring the transmitted x-ray intensity distribution at the detector input. The magnitude of this effect varies with both the apparent focal spot size and the magnification factor. It is well known that the apparent focal spot size increases toward the cathode side in the z -direction (anode to cathode direction) but remains unchanged in the transverse direction. The magnification factor, which dictates how much the focal spot is magnified when projected through a point object onto the detector, is determined by the object location and view angle. It is expected that due to the circular symmetry of cone beam reconstruction, the shapes of the PSFs do not vary with the angular position of the location in the x - y plane but they may vary with the z -level (coordinate) and the radial distance of the location (from the rotating axis).

To demonstrate the variation of the PSF shape with the z -level (coordinate) and the radial distance, the grayscale image of x' - y' , y' - z' , and x' - z' cross sections are plotted for three locations along the x -axis ($x=0, 6$, and 12 cm) and three different z -levels ($z=-5, 0$, and 5 cm, corresponding to $-4.8^\circ, 0^\circ, +4.8^\circ$ cone angle) in Figs. 6 and 7, respectively. The images in Fig. 6 show that the widths of the PSFs in the x' - (radial) and y' - (tangential) directions remained the same at different z -levels while those in z' -direction increased with the z -level. The spherical shaped PSF (at rotating center) is squeezed or elongated in the z' -direction to an ellipsoid-like PSF when it moved close to the anode or cathode. This is because the apparent focal spot size increases toward the cathode side in the z -direction but remains unchanged in the transverse direction.

The images in Fig. 7 show that the PSFs off the rotating axis became asymmetric and the shapes as well as the sizes of the PSFs were significantly changed with the radial distance from the rotation axis increasing. The off-axis PSFs have the shape of an elongated ellipsoid with the long axis tilting toward the rotating axis. The z -direction variation and tilting effect are probably the most unusual characteristics of the focal spot blurring effect in CBCT images.

III.D. Effects of combined detector and focal spot blurring

Since the detector blurring and focal spot blurring effects are equally important in typical imaging situations, the PSFs were computed for the combined detector blurring and focal spot blurring effects. Figures 6 and 7 were replotted with detector blurring added, as shown in Figs. 8 and 9, respectively. The PSF images in Figs. 8 and 9 look wider than those in Figs. 6 and 7, as expected due to the addition of the detector blurring effect. However, despite the larger blurring effects reflected by the generally wider PSFs, the shape and orientation variations with the radial distance and the z -level in Figs. 8 and 9 were similar to those with the focal spot blurring only PSFs in Figs. 6 and 7. Unlike the PSFs in Fig. 7, those in Fig. 9 had a more Gaussian-like shaped appearance due to the addition of the detector blurring.

The FWHMs in three major axes of the PSFs were computed and used to quantitatively demonstrate the size variation of the PSFs. Due to the tilting effect of the PSFs for off-axis locations, none of the major axes were along the radial or z directions. Therefore, we defined three major axis directions as tangential, pseudoradial and pseudo- z directions, the latter two of which tilted toward the rotating axis by a degree depending on the z -level and the radial distance. The FWHMs along the tangential, pseudoradial and pseudo- z directions were calculated for each PSF and plotted as a function of the z -level (Fig. 10) or radial distance (Fig. 11). Figure 10 shows that the FWHMs of the PSFs along the tangential and pseudoradial directions remained essentially the same at the five different z -levels, while those along the pseudo- z direction increased as the z -level increased. Figure 11 shows that the FWHMs of the PSFs in the pseudo- z direction increased more rapidly than those in the pseudoradial and tangential directions as the radial distance increased.

III.E. Effects of reconstruction filter

Since the Feldkamp reconstruction algorithm is mainly based on filtered backprojection, the filter selection may affect the spatial resolution properties of the reconstructed images. To better demonstrate the potential variation of the PSF with different filtering kernels, the FWHMs for five different reconstruction filters (Standard, Shepp–Logan, Cosine, Hamming, and Hanning) are plotted together for comparison in Fig. 12. Figure 12(a) shows the FWHMs at $z=5$ cm along the rotating axis. The FWHM values in the pseudoradial and tangential directions were identical (due to geometric symmetry) and gradually increased from left to right while that in the pseudo- z directions remained unchanged. This is because the filtration of the Feldkamp reconstruction was applied in the horizontal direction rather than the vertical direction. Figure 12(b) shows the FWHMs at $r=6$ cm. It is observed that the FWHM values in all directions gradually increased from left to right.

III.F. Comparison of simulated PSFs with measured PSFs

To validate the simulation method, two aluminum beads (one is located along rotating axis with $z=8.7$ cm, the other is located in rotating plane with $r=9.4$ cm) were imaged with an experimental benchtop CBCT system. The signal profiles of the bead were measured along the major axis through its center and compared with that computed from the simulated PSFs. This provides an indirect method to validate the simulation method. In Fig. 13, the two signal profiles are plotted together for comparison. The computed profiles agree well with the measured results, indicating that the simulation can be used to adequately represent the experimental imaging system used for the measurement. The FWHMs of computed profiles were 9.4% (at $z=8.7$ cm) and 8.5% (at $r=9.4$ cm) larger than those of measured results, respectively.

IV. DISCUSSION

Based on the results shown in Figs. 8-11, the effects of radial distance and cone angle on the PSFs in reconstructed cone beam CT images are both significant. Due to these effects, these PSFs are neither symmetric nor spatially invariant. Each of these PSFs is a function of 3D coordinates and can be considered as an ellipsoid with its density (PSF value) falling from the maximum at the center to zero at different rates along various radial directions. The PSFs in reconstructed cone beam CT images are asymmetric not only in that the ellipsoidal PSF function has different radii along its three major axes, but also in that the ellipsoid tends to tilt toward the rotating axis. This tilting effect may remind people of the tilting of the focal spot intensity profile when projected through an x-ray point hole camera onto the detector.

When the focal spot blurring effect dominates, the previous observations would become even more obvious, as shown by Figs. 6 and 7. When the detector blurring effect dominates, the PSFs would become symmetric, largely identical, and position independent. These indicate that the special characteristics of cone beam PSFs, such as asymmetry and spatially variant, come from the focal spot blurring rather than detector blurring. Although the PSFs may become asymmetric if the projection images are binned differently in horizontal and vertical directions, they are spatially invariant and oriented in the upright way. In actual imaging situations, the focal spot size and imaging geometry are usually selected to match the detector blurring characteristics so that the detector blurring and focal spot blurring effects are equally significant. Thus, the situation where either detector blurring or focal blurring dominates is probably not a realistic one.

Kwan *et al.*¹³ have recently reported on a series of MTF measurements for their cone beam breast CT system. It is logical to compare our simulation results with their experimental results. One fundamental difference between Kwan's study and ours is that they tried to measure the LSFs, while we tried to simulate true 3D PSFs. The two may produce different results because the LSF is actually an integral of the PSF along one or even two axes. Since Kwan's study included only one MTF measurement in the axial direction, many of our results cannot be compared with theirs, especially on the characteristics of the PSFs in the axial direction and their variation with the z -level (cone angle). For those that can be compared, the greatest difference lies in the variation of the spatial resolution in the x' - y' plane with the radial distance. Kwan's results show a substantial variation of the spatial resolution in the x' - y' plane with the radial distance, while ours show only a modest one. This is probably due to the fact that Kwan's measurement was performed on a system using continuous x-ray exposure during the scan. The relative motion between the detector and the scanned object would result in motion blurring whose magnitude increases linearly with the radial distance. Such motion blurring is not a problem for systems using pulsed exposures and was not included in our simulation. One example is with our experimental cone beam CT system which uses 10 ms (for regular exposure levels) or 30 ms (for higher exposure levels) exposures at a rate of 7.5/s for image acquisition. They would induce a motion blurring corresponding to ~7.5% or 22.5% of the angular increment between two consecutive views, respectively. This is substantially smaller than those induced by continuous exposure.

Our results may have several implications on the methods used to quantify the spatial resolution of cone beam CT images. The ellipsoidal shape of the 3D PSFs should imply that the PSFs should not be modeled as a spherically symmetric function. The tilting of the ellipsoidal PSFs toward the rotating axis indicates that even if the PSF is modeled as an ellipsoid, it should not be assumed to be oriented in the upright way. In fact, our simulation method may provide a way to predict the orientation of the PSF and may help model the

PSF more accurately in experimental measurement of the PSFs in cone beam CT. The variation of the PSF width in the axial direction implies that the use of a tilted thin metal wire to obtain oversampled LSF measurement with multiple slice images may be subject to a slight blurring effect due to the variation of the PSF along the axial and radial directions. However, if these images cover only a small distance, this blurring effect may not be significant to prevent the method from being used.

V. CONCLUSION

In this article the effects of the detector blurring, focal spot blurring, detector binning, and reconstruction filter on the PSF were studied. Our results show that the PSFs due to detector blurring are largely symmetric and vary little with the locations of the point objects. With focal spot blurring only or added to detector blurring, the PSFs along the rotation axis were largely symmetric but became increasingly asymmetric as the point objects were moved away from the rotation axis. The PSFs were found to become wider in the axial (anode to cathode) direction as the objects were moved toward the cathode side. The 3D PSFs may be approximated by an ellipsoid with three different axial lengths. They were found to point upright along the rotating axis but tilt toward the rotating axis as the point object was moved away from the axis.

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FIG. 1.
Flowchart for estimating the 3D PSFs.

FIG. 2.
Cone beam projection of a point object.

FIG. 3.
An illustration of projection of focal spot blurring.

FIG. 4. PSFs for detector blurring only plotted as grayscale images: (a) $x'-y'$ image at (0,0,0), (b) $x'-y'$ image at (12, 0, 5), (c) $y'-z'$ image at (0,0,0), and (d) $y'-z'$ image at (12,0, 5). All coordinates are in cm. Each image size is 0.73 mm $\times$ 0.73 mm.

FIG.5.

(a) x' - y' and (b) y' - z' images of the PSF for detector blurring only with 2×4 binning. Each image size is $0.73 \text{ mm} \times 0.73 \text{ mm}$.

FIG. 6. PSFs for focal spot blurring only plotted as grayscale images in the $x'-y'$, $y'-z'$, and $z'-x'$ planes. Each image size is $0.73 \text{ mm} \times 0.73 \text{ mm}$.

FIG. 7. PSFs for focal spot blurring only plotted as grayscale images in the $x'-y'$, $y'-z'$, and $z'-x'$ planes. Each image size is 0.73 mm x 0.73 mm.

FIG. 8. PSFs for combined detector and focal spot blurring plotted as grayscale images in the $x'-y'$, $y'-z'$, and $z'-x'$ planes. Each image size is $0.73 \text{ mm} \times 0.73 \text{ mm}$.

FIG. 9. PSFs for combined detector and focal spot blurring plotted as gray-scale images in the $x'-y'$, $y'-z'$ and $z'-x'$ planes. Each image size is 0.73 mm $\times$ 0.73 mm.

FIG. 10. FWHMs of PSFs (along pseudoradial, tangential, and pseudo-z directions) for combined detector and focal spot blurring plotted against different z-level. Two datasets (along and off the rotating axis, i.e., $r=0$ cm and $r=6$ cm) are plotted.

FIG. 11. FWHMs of PSFs (along pseudoradial, tangential, and pseudo-z directions) for combined detector and focal spot blurring plotted against different radial distance from the rotating axis. Two datasets (in and above the central plane, i.e., $z=0$ cm and $z=5$ cm) are plotted.

FIG. 12. FWHMs of PSFs at $z=5$ cm (a) and $r=6$ cm (b) reconstructed from five different reconstruction filters: Ramp, Shepp-Logan, Cosine, Hamming, and Hanning.

FIG. 13. CT number plotted along major axis for measured and computed images of two 0.8 mm diam aluminum beads at (a) $z=8.7$ cm and (b) $r=9.4$ cm.