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Model-based PSF and MTF estimation and validation from skeletal clinical CT images

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Purpose: A method was developed to correct for systematic errors in estimating the thickness of thin bones due to image blurring in CT images using bone interfaces to estimate the point-spread-function (PSF). This study validates the accuracy of the PSFs estimated using said method from various clinical CT images featuring cortical bones.

Methods: Gaussian PSFs, characterized by a different extent in the z (scan) direction than in the x and y directions were obtained using our method from 11 clinical CT scans of a cadaveric craniofacial skeleton. These PSFs were estimated for multiple combinations of scanning parameters and reconstruction methods. The actual PSF for each scan setting was measured using the slanted-slit technique within the image slice plane and the longitudinal axis. The Gaussian PSF and the corresponding modulation transfer function (MTF) are compared against the actual PSF and MTF for validation.

Results: The differences (errors) between the actual and estimated full-width half-max (FWHM) of the PSFs were 0.09 ± 0.05 and 0.14 ± 0.11 mm for the xy and z axes, respectively. The overall errors in the predicted frequencies measured at 75%, 50%, 25%, 10%, and 5% MTF levels were 0.06 ± 0.07 and 0.06 ± 0.04 cycles/mm for the xy and z axes, respectively. The accuracy of the estimates was dependent on whether they were reconstructed with a standard kernel (Toshiba’s FC68, mean error of 0.06 ± 0.05 mm, MTF mean error 0.02 ± 0.02 cycles/mm) or a high resolution bone kernel (Toshiba’s FC81, PSF FWHM error 0.12 ± 0.03 mm, MTF mean error 0.09 ± 0.08 cycles/mm).

Conclusions: The method is accurate in 3D for an image reconstructed using a standard reconstruction kernel, which conforms to the Gaussian PSF assumption but less accurate when using a high resolution bone kernel. The method is a practical and self-contained means of estimating the PSF in clinical CT images featuring cortical bones, without the need phantoms or any prior knowledge about the scanner-specific parameters. © 2014 American Association of Physicists in Medicine. [http://dx.doi.org/10.1118/1.4835515]

Key words: clinical CT, resolution, 3D PSF, MTF, PSF measurement

1. INTRODUCTION

The point-spread-function (PSF) and the modulation transfer function (MTF) of a clinical CT system provide fundamental information regarding its spatial resolution. The PSF for a given image data set is a function of proprietary reconstruction kernels and other variables that are highly specific to each manufacturer. Clinical CT users do not have direct access to image PSFs. Users are simply guided by an inexplicit knowledge regarding the effect of reconstruction kernels and other parameter choices on the final image quality. Even with exact knowledge of the PSF of the CT system within slice planes, PSF determination along the longitudinal axis is further complicated by dependence on the scanning mode (i.e., helical, volume, or cone-beam) and manufacturer specific computational implementation.

In CT imaging, the PSF has clinical relevance beyond its role as a periodic quality assurance metric for the system. The PSF determines smallest feature sizes which are accurately captured in a CT image set, and hence the degree of confidence in the true geometry and intensity of an object’s image. One notable example is for measuring the thickness and density of cortical bones from CT images, which has implications for predicting fracture risk. Dougherty et al.
determined that cortical structures smaller than 1.5–2.0 times the full-width half-max (FWHM) of the PSF will be subject to significant thickness underestimation and density underestimation due to blurring. In addition, CT based patient-specific finite element models and implant manufacturing are also highly sensitive to the accuracy of the segmentation of cortical bone features. The PSF, if known, may also be used in a deconvolution algorithm to restore or enhance the spatial resolution in three dimensions.

In this study, a model-based method previously developed for purpose of estimating the thickness and intensity of cortical bones is extended towards estimation of the PSF directly from any clinical CT image. The primary requisite of this approach is that: (a) the scanned region contains cortical bone features, (b) the PSF can be approximated to be Gaussian-shaped, and (c) the PSF can be reasonably simplified to be separable and shift-invariant in all dimensions. The accuracy of this method is validated by comparing the resulting estimated PSF and MTF values generated from the CT image of a cadaveric craniofacial specimen to the directly measured PSF and MTF from a clinical CT scanner.

2. BACKGROUND

2.A. Gaussian model for blurring of bone in CT

Previously, a blurring model for thin cortical bones in clinical CT images has been proposed. What follows is a brief summary and the extension of the previous work towards estimating the PSF.

Let the nonblurred image of a thin cortical bone be represented by a rectangular function

\[ f(x_1, x_2) = Y \cdot \Pi \left[ \frac{x_1 \cos \theta - x_2 \sin \theta}{W} \right], \]  

where it is perpendicular to the plane spanned by \( \{x_1, x_2\} \), and rotated \( \theta \) degrees with respect to the \( x_2 \) axis (Fig. 1). The width of the cortical bone is \( W \) and its intensity is \( Y \). For mathematical simplicity \( f(x_1, x_2) \) is centered on the origin.

It has been widely adopted in literature that the blurring process in CT scanners can be approximated by convolution of the ideal nonblurred image with a Gaussian function. Furthermore, for simplicity it is assumed that the point-spread-function (PSF) is reasonably approximated to exhibit radial symmetry within the slice plane and shift-invariance in all dimensions. As such, the PSF of the imaging system can be modeled as a normalized Gaussian function given by

\[ h(\bar{x}) = \frac{1}{(2\pi)^{W/2}|\Sigma|^{1/2}} e^{-\frac{\bar{x}'^T \Sigma^{-1} \bar{x}'}{2}}, \]

where \( \bar{x} \in \mathbb{R}^{N=2} = (x_1, x_2) \) and \( \Sigma \in \mathbb{R}^{N \times N=2 \times 2} = \begin{bmatrix} \sigma_1^2 & 0 \\ 0 & \sigma_2^2 \end{bmatrix} \). The diagonal covariance matrix is chosen with the assumption that the blurring in the \( x_1 \) and \( x_2 \) directions are independent. The blurred CT image of the cortical bone is the convolution of Eqs. (1) and (2). Let \( g(x_1', x_2') \) be the blurred image in a rotated coordinate system:

\[ g(x_1', x_2') = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} Y \cdot \Pi \left[ \frac{x_1' \cos \theta - x_2' \sin \theta}{W} \right] \times \frac{1}{2\pi |\Sigma|^{1/2}} e^{-\frac{(x_1 - \bar{x})^T \Sigma^{-1} (x_1 - \bar{x})}{2}} \, dx_1 \, dx_2, \]

(3)

where \( \bar{x} = (r_1, r_2) \), \( U \) is a transform matrix which applies a counterclockwise rotation by \( \theta \) degrees and \( \bar{x}' = U \bar{x} \). It can be shown that

\[ g(x_1', x_2') = \frac{Y}{\sqrt{2\pi}\sigma_r} \int_{-\infty}^{\infty} \frac{e^{-\frac{\bar{x}'^2 - \bar{x}^2}{2\sigma_r^2}}}{\sigma_r} \, d\tau_1 \]

\[ = \frac{Y}{\sqrt{2}} \left[ \text{erf} \left( \frac{x_1' + \frac{W}{2 \sigma_r}}{\sigma_r \sqrt{2}} \right) - \text{erf} \left( \frac{x_1' - \frac{W}{2 \sigma_r}}{\sigma_r \sqrt{2}} \right) \right], \]

(4)

where \( \text{erf}(\cdot) \) is the error function and

\[ \sigma_r = \sqrt{\sigma_1^2 \cos^2 \theta + \sigma_2^2 \sin^2 \theta}. \]

From herein, axes spanning the transverse plane of CT scanner are denoted as the \( x \) and \( y \) axes, and the longitudinal axis as the \( z \) axis. For the case where the normal of the cortical bone surface lies within the CT image slice plane, let \( \sigma_1 = \sigma_x \) and \( \sigma_2 = \sigma_y \). By approximating the blurring to be isotropic within the slice \( (\sigma_x = \sigma_y) \), then \( \sigma_r \) is constant for all \( \theta \)—which can be set to define the in-plane component \( \sigma_{xy} \). The blurring in the \( z \) axis is different compared to the blurring within the slice plane. Therefore, for the case of a cortical bone with a surface normal parallel to one of the other two planes orthogonal to the \( xy \) plane, the blurring is a function of the rotation angle \( \theta \), \( \sigma_1 = \sigma_{xy} \), and the longitudinal component \( \sigma_z = \sigma_z \).
FIG. 2. Idealized illustrations of two generic types of intensity profiles extracted from a CFS CT image (top). The profile from the posterior maxillary sinus wall and the zygomatic buttress may be represented by \( n = 3 \) (bottom left) and \( n = 5 \) (bottom right) according to Eq. (6) (black plots). The rectangular functions representing the true intensity and geometry of the nonblurred layers are also shown (grey plots).

2.B. Generalized model for blurring of cortical features in CT

The more general case may be considered, where the bone is modeled to include adjacent layers of variable mineral density, surrounded on the outside by either air or soft-tissue. The craniofacial skeleton (CFS) is a unique anatomical region where the shell bones are entirely cortical, or have a bicortical structure in which a trabecular region is sandwiched between high density cortical layers. With reference to Fig. 2, by inclusion of the additional layers as a summation of rectangular functions, and substituting the integration bounds in Eq. (4), the blurred profile is given by

\[
g(x_i') = \frac{Y_1}{2} \left[ \text{erf} \left( \frac{x_i' - X_0}{\sigma r \sqrt{2}} \right) - \text{erf} \left( \frac{x_i' - X_1}{\sigma r \sqrt{2}} \right) \right] \\
+ \frac{Y_n}{2} \left[ \text{erf} \left( \frac{x_i' - X_n}{\sigma r \sqrt{2}} \right) - \text{erf} \left( \frac{x_i' - X_f}{\sigma r \sqrt{2}} \right) \right] \\
+ \frac{1}{2} \sum_{j=2}^{n-1} Y_j \left[ \text{erf} \left( \frac{x_i' - X_{j-1}}{\sigma r \sqrt{2}} \right) - \text{erf} \left( \frac{x_i' - X_j}{\sigma r \sqrt{2}} \right) \right].
\]

(6)

where \( n \) is the number of layers, \( X_i \) are the boundaries of each layer, and \( Y_i \) are the unblurred intensity of the layer. Equation (6) represents a generalized mathematical model of a blurred profile obtained by mapping intensity values onto a line that perpendicularly traverses the bone mineral layers of interest. For a single cortical layer and a bicortical cross-section, \( n = 3 \) and \( n = 5 \), respectively (Fig. 2). Higher values of \( n \) may be used for additional layers.

In a previous study, the process of estimating the set of unknown values \( \{\sigma r, X_1 \ldots, X_{n-1}, Y_2 \ldots, Y_{n-1}\} \) in Eq. (6) using a nonlinear optimization solver is outlined in detail. To achieve stable solutions, the variables are explicitly bounded within conservative, but anatomically relevant windows; for example, the \( Y \) variables representing peak intensities for cortical and trabecular bone may only fall within 800–1800 and 100–800 HU, respectively. Similarly, the thicknesses for craniofacial cortical bone are within 0.2–3 mm. The validation study using a cadaveric CFS specimen demonstrated good agreement between the estimates and the true intensity and thickness measurements. While \( \sigma r \) was among the parameters estimated in the previous investigation, it could not be dimensionally decomposed into its within-slice and axial components as estimation of the PSF without the developments presented in the current study.

Depending on the selected acquisition mode, blurring may be nearly shift invariant or vary gradually throughout the acquired volume. In cases where blurring varies spatially, the blurring kernel determined from a bone interface is suitable to correct bone thicknesses in the vicinity of said interface and a different kernel would be needed if the method were applied elsewhere in the volume. Thus, the mathematical requirement
is for the blurring to be shift invariant only over small dimensions comparable to the kernel size so that convolution may be used. Radial symmetry of the PSF in the $xy$ plane is a reflection of the circular symmetry of the acquisition geometry.

3. METHODS

3.A. Cortical blurring model-based PSF estimation

In this study, the $\sigma_r$ parameter of the PSF is decomposed into $\sigma_{xy}$ and $\sigma_z$ to estimate the three-dimensional PSF. The parameter $\sigma_{xy}$ is first estimated based on intensity profiles extracted normal to the surface of shell bones that are aligned perpendicular to the image slice. Subsequently, $\sigma_r$ is estimated from bones that are aligned perpendicular to the $xz$ or $yz$ planes, from which $\sigma_z$ is calculated by inputting $\sigma_{xy}$ and the profile angle $\theta$ in Eq. (6). From herein, this method of estimating the PSF using the proposed blurring model will be referred to as Blurring Model Point-Spread Estimation (BMPE).

A fresh-frozen cadaveric head was scanned on a Toshiba Aquilion ONE (Toshiba Medical Systems, Otawara, Japan), using 11 combinations of scan parameters as listed in Table I. This scanner features 320 detector rows, which captures image slices in a conventional helical scan mode, as well as a volume mode where up to 320 parallel image slices can be captured simultaneously with one rotation of the gantry. The reconstruction kernels FC68 and FC81 refer to “standard” and the high resolution “bone” filters, respectively. The x-ray tube currents were varied between 200 and 500 mA. The Aquilion ONE scanner utilizes two spot sizes: $1.6 \times 1.4$ mm for high currents such as 400 and 500 mA used for image sets 8–11 and $0.9 \times 0.8$ mm used for lower currents such as image sets 6 and 7. The slice thickness was varied between 0.3 and 1.5 mm. All scans were performed at a tube voltage of 120 kVp. This scanner features Toshiba’s Adaptive Iterative Dose Reduction (AIDR 3D) technology, which was enabled by default for all the reconstructions.

To estimate the three-dimensional PSF, a total of 12 intensity profiles were used for each CT image set; six profiles to estimate $\sigma_{xy}$, and six to estimate $\sigma_z$. The profiles were extracted from shell bones with distinct cortical boundaries; for example, from the walls of maxillary and frontal sinuses, and the vomer, nasal, temporal, or zygomatic buttress and arch bones depending on the plane of interest (Fig. 3). Perpendicular bones were selected by observing the evolution of the bone of interest over at least five slices, and only those bones that appeared to be stable in geometry and intensity were used. This criterion also ensures the cortical walls have minimal curvature. The process of manual profile selection and the optimization computation was repeated three to four times per image set to calculate the average and standard deviation of the $\sigma$ values estimates.

3.B. Validation of PSF estimations in three dimensions

The PSF of the scanner within the slice plane and the longitudinal axis were measured by determining the line-spread-functions (LSFs) from the angled slit method as outlined by Boone et al. This method is a practical approach for sampling the LSF at a significantly higher rate than the Nyquist frequency in order to construct the presampled PSF/MTF of a CT image. From herein, the term PSF will continue to be used in place of the LSF measured from the slit phantom images.

---

**Table I. Summary of scan parameters used for the validation study.**

<table>
<thead>
<tr>
<th>Image set</th>
<th>X (mm)</th>
<th>Y (mm)</th>
<th>Z (mm)</th>
<th>Scan mode</th>
<th>Kernel</th>
<th>mA</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.43</td>
<td>0.43</td>
<td>1</td>
<td>Helical</td>
<td>FC68</td>
<td>300</td>
</tr>
<tr>
<td>2</td>
<td>0.43</td>
<td>0.43</td>
<td>0.3</td>
<td>Helical</td>
<td>FC68</td>
<td>300</td>
</tr>
<tr>
<td>3</td>
<td>0.43</td>
<td>0.43</td>
<td>0.5</td>
<td>Volume</td>
<td>FC68</td>
<td>200</td>
</tr>
<tr>
<td>4</td>
<td>0.43</td>
<td>0.43</td>
<td>0.5</td>
<td>Volume</td>
<td>FC68</td>
<td>400</td>
</tr>
<tr>
<td>5</td>
<td>0.43</td>
<td>0.43</td>
<td>0.5</td>
<td>Volume</td>
<td>FC68</td>
<td>500</td>
</tr>
<tr>
<td>6</td>
<td>0.43</td>
<td>0.43</td>
<td>1.5</td>
<td>Volume</td>
<td>FC81</td>
<td>200</td>
</tr>
<tr>
<td>7</td>
<td>0.43</td>
<td>0.43</td>
<td>0.5</td>
<td>Volume</td>
<td>FC81</td>
<td>200</td>
</tr>
<tr>
<td>8</td>
<td>0.43</td>
<td>0.43</td>
<td>1.5</td>
<td>Volume</td>
<td>FC81</td>
<td>400</td>
</tr>
<tr>
<td>9</td>
<td>0.43</td>
<td>0.43</td>
<td>0.5</td>
<td>Volume</td>
<td>FC81</td>
<td>400</td>
</tr>
<tr>
<td>10</td>
<td>0.43</td>
<td>0.43</td>
<td>1.5</td>
<td>Volume</td>
<td>FC81</td>
<td>500</td>
</tr>
<tr>
<td>11</td>
<td>0.43</td>
<td>0.43</td>
<td>0.5</td>
<td>Volume</td>
<td>FC81</td>
<td>500</td>
</tr>
</tbody>
</table>

**FIG. 3.** Examples of regions where profiles were taken from the cadaveric CFS CT images for the BMPE method from shell bones with distinct cortical boundaries such as the walls of maxillary and frontal sinuses, and the vomer, nasal, temporal, or zygomatic buttress and arch bones on each of the three primary orthogonal viewing cross-sectional planes ($xy$, $xz$, and $yz$).
A 60 μm thick aluminum foil was sandwiched between two slabs of water equivalent polymer. The phantom was scanned twice for each of the scan settings listed in Table I: once oriented perpendicular to the xy axis and again oriented perpendicular to the xz/yz axes. A total of six image slices from each scan were cropped down to provide approximately perpendicular to the xz/yz axes. A total of six image slices once oriented perpendicular to the xy axis and again oriented perpendicular to the xz/yz axes. A total of six image slices from each scan were cropped down to provide approximately 105 PSF profile samples each. The slit angle was varied between 2° and 10°. This variation was accounted for in the processing steps by precisely ascertaining the “half-pixel-shifted alignments” in order to calculate the phase shift between profile samples28,29 (Fig. 4). This is in contrast to the visually compensated method suggested by Boone et al. From these, six composite LSF curves were calculated and then averaged to be used as the measured PSF. As per Boone et al., a noise removal step was performed by convolving the curves with a rectangular window twice. The width of this window was varied between 2 and 20 sampling units, as the higher values were necessary for exceptionally noisy PSFs such as in the longitudinal axis of image set 10. The trailing end tails of the composite PSFs were exponentially extrapolated to reduce truncation error and to smoothen the MTF curves at frequencies approaching zero. The MTFs were calculated by taking the Fourier transform of the corresponding PSFs, and the final results were divided by the appropriate Sinc function to compensate for the finite width of the slit (60 μm); however, the differences before and after this compensation were not discernible in the MTF values.

The FWHM of each PSF measured from the slit phantoms were compared to the FWHM of the estimated PSFs determined by the BMPE method. The frequency values corresponding to MTF levels of 75%, 50%, 25%, 10%, and 5% contrast were also obtained and compared. Welch’s t-tests were to test for significance differences in errors.

4. RESULTS

Figures 5(a) and 5(b) illustrate typical PSF shapes corresponding to the FC68 (set 5) and FC81 (set 10) reconstruction kernels, respectively, along with the corresponding PSFs derived from the BMPE method. Similarly, Figs. 5(c) and 5(d) illustrate the MTF calculations from image sets 4 and 6. The overall results of FWHM values for the PSFs are summarized in Table II for all image sets within the plane of slices (xy axes) and along the longitudinal z axis. The overall (systematic) differences (absolute error) between the measured and estimated PSFs were 0.09 ± 0.05 and 0.14 ± 0.11 mm for the xy and z axes, respectively. The overall errors in the predicted frequencies at the chosen MTF levels were 0.06 ± 0.07 and 0.06 ± 0.04 cycles/mm for the xy and z axes, respectively.

The FWHM value for image sets 1–5 (FC68 kernel) have a mean error of 0.06 ± 0.05 mm, while in sets 6–11 (FC81 kernel) this mean error increases to 0.12 ± 0.03 mm. Similarly the mean error in the estimated MTFs in the FC68 groups is 0.02 ± 0.02, compared to 0.09 ± 0.08 cycles/mm in the FC81 group. The MTF values at frequencies approaching the cutoff make the greatest contribution to this error; in the FC81 group, the MTF mean error at 75%, 50%, 25%, and 10% contrast level is 0.06 ± 0.06 cycles/mm, while at the 5% mark it increases to 0.21 ± 0.03 cycles/mm.

5. DISCUSSION

The resolution information provided by a PSF of the CT image provides valuable and clinically relevant insight, such as the smallest cortical or trabecular bone geometry and intensity information which may be accurately discerned in an image. While the slant-slit method is a relatively practical means of measuring the MTF for research purposes, the procedure is still far from trivial to enable its use routinely, particularly in a clinical setting. This method and other proposed simplified approaches require a phantom to be scanned separately along with each scan on a patient, as well as numerous postacquisition processing steps in order to arrive at the PSF for a given set of scan parameters. To complicate matters further, modern scanners may utilize adaptive reconstruction algorithms for dose reduction and image enhancements which result in a subject-dependent PSF. This may result in significantly different PSFs between a phantom and a clinical scan performed at identical technique settings.

The approach presented here allows users to estimate the PSF of the CT image directly without addition scanning or modification of protocols to include reference markers. This method can be used on any previously-acquired CT image data or from those sourced from a clinical scanner that may not be directly accessible.

The MTF of a CT image is a detailed characterization of its resolution; it is dependent on inherent properties of the scanner such as the detector size, focal spot size, as well as the reconstruction filter choice which modulates the trade-off between noise and spatial resolution. A validation study by Nickoloff et al. has previously confirmed the accuracy of Gaussian approximation for the within-slice PSF in previous generations of CT scanners, although reconstruction
methods with “edge enhancement” make this model less reliable.22 The PSFs within the slice planes in this study also support this conclusion: the FWHM value for the FC68 kernel sets have a lower mean error than the FC81 kernel image sets. Comparing the MTFs, the frequencies approaching the cutoff frequency make the greatest contribution to the differences in the estimation errors. In the FC68 group, there is no statistically significant difference in the estimation errors at the various MTF levels. Note that while this agrees with the expectation that a Gaussian estimate is most likely unable to capture the high-pass characteristics of the FC81 kernel, it is not entirely clear to what degree the inherently higher noise levels in the slit phantom images of this group might be contributing to the estimation errors close to the cutoff frequency.

### Table II. The overall PSF results comparing the FWHM of values measured from slit-phantom images to the estimates from the BMPE method. The errors values which indicate a statistically significant difference (p ≤ 0.01) are marked with *.

<table>
<thead>
<tr>
<th>Image</th>
<th>CT slice plane (xy)</th>
<th>Longitudinal axis (z)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Measured</td>
<td>Estimated</td>
</tr>
<tr>
<td>Set 1</td>
<td>1.56 ± 0.06</td>
<td>1.44 ± 0.06</td>
</tr>
<tr>
<td>Set 2</td>
<td>1.44 ± 0.12</td>
<td>1.48 ± 0.07</td>
</tr>
<tr>
<td>Set 3</td>
<td>1.44 ± 0.07</td>
<td>1.44 ± 0.07</td>
</tr>
<tr>
<td>Set 4</td>
<td>1.51 ± 0.05</td>
<td>1.48 ± 0.07</td>
</tr>
<tr>
<td>Set 5</td>
<td>1.53 ± 0.05</td>
<td>1.41 ± 0.07</td>
</tr>
<tr>
<td>Set 6</td>
<td>0.79 ± 0.01</td>
<td>0.71 ± 0.06</td>
</tr>
<tr>
<td>Set 7</td>
<td>0.79 ± 0.07</td>
<td>0.71 ± 0.06</td>
</tr>
<tr>
<td>Set 8</td>
<td>0.97 ± 0.00</td>
<td>0.82 ± 0.06</td>
</tr>
<tr>
<td>Set 9</td>
<td>0.95 ± 0.01</td>
<td>0.82 ± 0.06</td>
</tr>
<tr>
<td>Set 10</td>
<td>0.92 ± 0.04</td>
<td>0.87 ± 0.06</td>
</tr>
<tr>
<td>Set 11</td>
<td>0.98 ± 0.04</td>
<td>0.82 ± 0.06</td>
</tr>
</tbody>
</table>

Note: Dimensions are in mm.
While the BMPE method has reduced accuracy with the FC81 group of images, it is interesting to note that it exhibits enough precision to detect the differences in the FWHM size of PSF due to focal spot size changes (in the image sets with >400 mA setting versus lower ones). The PSF measured with the slit phantom images reconstructed with FC81 are significantly wider at the larger focal spot sizes (mean difference of 0.18 mm, p < 0.01). The difference measured by applying BMPE with the same scan settings is 0.13 mm (p < 0.01). For image sets reconstructed with FC68, neither the slit phantom measurements nor the BMPE method could detect a discernible difference in PSF FWHM due to changes in focal spot size.

The measured 3D PSF of a CT scanner does exhibit shift-variance and it is not generally separable. The inherent simplifications in the BMPE method does not account for variability. This is a primary limitation of this method, in addition to the deviations from the Gaussian assumption of the kernels described above. In a cone-beam scanner this may have greater implications since the PSF varies gradually as the distance increases from the central fan beam. Therefore, the PSF estimates are more likely to be accurate if the cortical profiles are confined to a small region of interest. Furthermore, with reference to Eq. (5), the $\sigma_z$ parameter estimate using the BMPE method is dependent on the previously estimated $\sigma_0$ value. Therefore, it partially inherits any errors from that dimension as well. PSF estimates limited to a small volume of interest are more likely to produce more accurate results.

This study has validated that the PSF and MTF curves computed using the BMPE method are highly accurate in all dimensions for a standard reconstruction kernel, such as the FC68, and also reliable for a high resolution kernel such as the FC81 at frequencies attenuated between 0% and 90%. In the case of specialized reconstruction kernels such as FC81, the shape of the measured PSF deviates from a Gaussian shape, and the BMPE method underestimates the PSF FWHM. However, it is conceivable that the PSF shape assumption may be adapted to non-Gaussian shapes. As discussed in the original introduction of this method, the only information required to effectively apply it are anatomical knowledge of the region for most ideal selection of thin-cortical features, such as the general range of HU intensity values expected in the cortical bone.

It is also instructive to discuss the specific reconstruction method used by The Toshiba Aquilion ONE in this study. This scanner uses a method dubbed “AIDR 3D” which allows the scans to be acquired at lower dose while maintaining a similar appearance. Using this method, a blurred image and an edge enhanced image are added together according to weighting factors $k$ and $k-1$, respectively, where $k \in [0, 1]$ varies at each voxel. The value of $k$ is determined by an “edge calibration algorithm” to increase the contribution of the blurred images in low contrast parts of the image (high $k$) and increase the contribution of the enhanced image at edges (low $k$). Thus, the method is called “edge preserving” although depending on the choice of filters and weighting factors, some edge enhancing may occur.

A study by Gervaise et al. has indicated that there is no perceived change in resolution in clinical images whereas there is noticeable noise reduction in low contrast regions when using AIDR 3D. This finding is expected because AIDR 3D is used in conjunction with less than twofold dose reduction to generate “familiar looking” images that hide increased x-ray noise. Little edge enhancement can be achieved without making the increased x-ray noise noticeable. Thus we expect edges which are the features of interest in determining bone thickness to be less affected by AIDR 3D processing. The bone edges are objects having similar high contrast to noise therefore limiting the range of $k$ values that affect image processing over the image. As a result, we expect an edge blurring estimate obtained at particular location to be at the very least applicable to a neighborhood where the estimate was obtained. It is also expected that edge images acquired from different scans with similar x-ray technique factors will have similar profiles.

6. CONCLUSIONS

This study validated a method for estimating the 3D PSF and MTF retrospectively from clinical CT images, without the need for additional imaging. It is proposed that it may be universally applied to images of any bony anatomy that features distinct cortical walls, so long as the PSF may be reasonably assumed to be Gaussian-shaped and shift-invariant in three dimensions. The accuracy of this method has been demonstrated by comparing the resulting estimated PSFs and MTFs from CT images of a cadaveric craniofacial skeleton to the directly measured PSFs and MTFs from phantoms on a clinical CT scanner using multiple scan parameters, including different reconstruction kernels.

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