Calibration-free device sizing using an inverse geometry x-ray system

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Purpose: Quantitative coronary angiography (QCA) can be used to support device size selection for cardiovascular interventions. The accuracy of QCA measurements using conventional x-ray fluoroscopy depends on proper calibration using a reference object and avoiding vessel foreshortening. The authors have developed a novel interventional device sizing method using the inverse geometry scanning-beam digital x-ray (SBDX) fluoroscopy system. The proposed method can measure the diameter and length of vessel segments without imaging a reference object and when vessels appear foreshortened.

Methods: SBDX creates multiple tomosynthetic x-ray images corresponding to planes through the patient volume. The structures that lie in the plane are in focus and the features above and below the plane are blurred. Three-dimensional localization of the vessel edges was performed by examining the degree of blurring at each image plane. A 3D vessel centerline was created and used to determine vessel magnification and angulation relative to the image planes. Diameter measurements were performed using a model-based method and length measurements were calculated from the 3D centerline. Phantom validation was performed by measuring the diameter and length of vessel segments with nominal diameters ranging from 0.5 to 2.8 mm and nominal lengths of 42 mm. The phantoms were imaged at a range of positions between the source and the detector (± 16 cm relative to isocenter) and with a range of foreshortening angles (0° – 75°).

Results: Changes in vessel phantom position created magnifications ranging from 87% to 118% relative to isocenter magnification. Average diameter errors were less than 0.15 mm. Average length measurements were within 1% (0.3 mm) of the true length. No trends were observed between measurement accuracy and magnification. Changes in vessel phantom orientation resulted in decreased apparent length down to 28% of the original nonforeshortened length. Average diameter errors were less than 0.25 mm across all vessel angulations; errors were less than 0.1 mm for smaller diameter vessels and low to moderate vessel angles. Diameter errors increased with true diameter and vessel angle relative to the image plane. Average length measurement errors were also within 1% (0.3 mm) for each angulation.

Conclusions: Tomosynthetic imaging with SBDX can accurately measure dimensions of vessels in various magnifications and angulations without calibration. This method may be more accurate and convenient than conventional QCA techniques. © 2011 American Association of Physicists in Medicine. [DOI: [10.1118/1.3528227](https://doi.org/10.1118/1.3528227)]

Key words: quantitative coronary angiography, inverse geometry, fluoroscopy, tomosynthesis, angiography

I. INTRODUCTION

Balloon angioplasty and stent placement are two frequent interventions performed in catheterization laboratories to treat cardiovascular disease. Clinical outcomes are directly impacted by the clinician's ability to choose the correct device given the size of the lesion and surrounding healthy vessel (Fig. 1). Underdeployed devices increase the risk of

(1) acute closure due to stent thrombosis^{1,2} and (2) gradual restenosis with recurrence of anginal symptoms.^{3,4} Oversized devices may cause dissection, leading to acute closure or coronary perforation.^{4–6} Devices that are too short with respect to the lesion can result in edge dissection,⁷ restenosis,⁸ and unanticipated use of more devices. Overlapping multiple stents can also lead to restenosis.^{9,10} Excessively long devices risk thrombosis¹¹ and restenosis.^{12–14} Choosing the cor-

FIG. 1. Measuring vessel dimensions from coronary angiograms. In order to correctly select an interventional device to treat a stenosis, the physical dimensions of the vessel must be determined. The angioplasty balloon or stent should match the (a) diameter of the healthy vessel and (b) length of the lesion.

rect device improves patient outcomes, avoids additional interventions, and reduces excess device utilization, which can be costly.

Intravascular ultrasound (IVUS) is the current gold standard for measuring lesions during and after an intervention.^{15,16} High-resolution 2D tomographic imaging with motorized pullback can accurately characterize the vessel cross section and lesion eccentricity over the length of the vessel. Despite these imaging capabilities, IVUS is not routinely used because of the added cost of the imaging catheters and additional procedure time. Risks of IVUS imaging include vessel spasm, vessel dissection, and damage to deployed stents.¹⁷ Accordingly, interventionalists must weigh the benefits of IVUS against the costs and risks to determine when it is warranted.

The main alternative to IVUS for interventional sizing of vessels is quantitative coronary angiography (QCA), which measures vessel dimensions from x-ray angiograms. Though widely available, this technique is not frequently used, presumably because of the time required to obtain accurate results. A key component to accurate QCA is the calibration step required to convert image measurements to physical dimensions.¹⁸ In x-ray imaging with a point source and large area detector, all anatomy is projected onto a plane and the magnification of each point on a vessel must be determined to accurately convert image pixel measurements to physical vessel dimensions. Imaging an object of known size in the same field-of-view provides a reference to determine the magnification of the angiogram. However, an extra step to analyze the reference object is required, any error in the measurement of the reference object carries over to vessel measurements, and the calibration is valid only in the region around the reference object. Another weakness of conventional QCA is the potential of foreshortening of vessels in the image. Vessels that are not parallel to the image plane will appear shortened by the projection imaging process and spe-

cial attention must be paid to the angulation of the gantry when recording the images. In practice, most interventionalists make an educated guess of vessel diameter and lesion length instead of using quantitative measurements.

Scanning-beam digital x-ray (SBDX) is a low dose inverse geometry fluoroscopy system capable of reconstructing multiple tomosynthetic images in real time,¹⁹ which may overcome some of the shortcomings of conventional QCA. Tomosynthetic reconstruction creates an image corresponding to a physical plane, where structures located in the plane are in focus and structures above and below the plane are blurred.²⁰ We have developed a novel algorithm for exploiting the properties of SBDX tomosynthetic images for performing QCA. This method can determine absolute vessel diameters without imaging a reference object. In addition, it can obtain accurate length measurements when vessels would appear foreshortened in a 2D x-ray image. In this paper, we describe our device sizing algorithm and phantom studies demonstrating feasibility.

II. BACKGROUND

II.A. SBDX

The SBDX imaging system geometry and image formation method have been previously described in detail.¹⁹ Briefly, the system uses an electron beam scanned over a large area transmission target with a prepatient multihole collimator. The beams exit the collimator holes and strike the small area, solid state, photon counting detector, creating a series of overlapping narrow field-of-view images from a range of focal spot positions (up to 100×100). Shift-and-add tomosynthesis is used to reconstruct the full scanning field-of-view at a set of planes for each scan frame (up to 30 scans/s). Image plane locations are independently programmable, with up to 16 planes reconstructed simultaneously in real time by dedicated hardware. The planes are parallel to the source and detector planes. To reconstruct a single image plane, the tomosynthesis algorithm first defines the boundaries for each pixel in the plane. The detector images are back-projected and summed at the image plane. The contribution of each back-projected detector element to a reconstructed image pixel is based on the area of overlap. To create an image similar to a conventional system, in which all structures are in focus, a multiplane composite image is formed from a set of single-plane images. The multiplane composite is generated by a plane selection algorithm in which first the local contrast and sharpness is evaluated at each pixel in each plane and then, on a pixel-by-pixel basis, the pixel from the plane of highest focus is selected for display (Fig. 2).

The 3D spatial coordinates of each reconstructed pixel in each plane are precisely known relative to the system gantry. Given a set of tomosynthetic images spanning a volume, an object can be localized in 3D based on the degree of object focus in each image plane. The feasibility of localizing small, high contrast objects has been demonstrated in phantoms with submillimeter accuracy.²¹ The same 3D object localization technique can be applied to localize an edge of a

Fig. 2. Tomographic blur of single-plane images and multiplane composite. (a) Schematic showing a vessel segment (gray) spanning several image planes (dotted lines). (b) In each of the single-plane images, the part of the vessel near the image plane is in focus, while portions above and below are blurred. (c) A set of single-plane images is combined into a multiplane composite, where all portions of the vessel appear in focus, similar to conventional fluoroscopy. Note that the curve in the vessel phantom wire is physical and is not an imaging artifact. For clarity, the system dimensions have been exaggerated.

larger object by restricting the analysis to a volume containing the edge and treating the edge as a discrete object. The sizing algorithm described in this article localizes the edges of a contrast filled vessel in a piecewise fashion to determine the 3D centerline of the vessel. The 3D centerline can be used in QCA to compute the length of a vessel segment and the magnification of the vessel.

II.B. Quantitative coronary angiography

QCA yields relative and absolute dimensions of coronary vessels (diameter, cross sectional area, percent narrowing, and lesion length), which can be used for both diagnosis and interventional decision making. QCA algorithms typically determine diameter by detecting vessel edges. Edge detection methods can be divided into two approaches: (1) Derivative-based, which uses first and second derivatives of the profiles to determine vessel boundaries, and (2) model-based, which assumes a parametrized vessel profile model and finds a maximum likelihood solution that best matches the observed profile.²² The algorithm described in this paper uses a model-based approach which will be described in Sec. II C.

Note that conventional QCA methods require a calibration step to produce measurements in physical units [e.g., millimeters (mm)] rather than pixels. Several strategies have been proposed,¹⁸ the most common of which is to image a contrast filled catheter near the vessel of interest. For example, dividing the true inner diameter of the catheter lumen by the apparent diameter, in pixels, produces the necessary scaling factor to convert pixel measurements to physical units. However, any error in this step propagates to the vessel measurements.

II.C. Model-based QCA

Model-based methods assume that the cross sectional profile corresponds to that of a generalized cylinder, and the parameters of a multiparameter model are jointly optimized to find the set that yields the best fit between the observed and modeled profile in a least-squares sense. Thorough explorations of model-based vessel measurements are provided by Pappas and Lim²² and Chan *et al.*²³ Their presentation is adapted here for the purpose of working with intensity images with dark vessels on a bright background. The overall model for an observed profile $I(t)$ is summarized by

$$I(t) = I_b(t) - [I_v(t) * g(t)] + I_n(t), \quad (1)$$

where t is the position along the length of the profile. The observed profile $I(t)$ is the result of the background $I_b(t)$ minus the ideal vessel profile $I_v(t)$ convolved with the system line spread function (LSF) $g(t)$. The function $I_n(t)$ represents the noise component.

We model the ideal vessel profile by assuming the signal difference profile generated by a contrast-enhanced vessel is the projection of a cylinder

$$I_v(t) = \begin{cases} 2\alpha\sqrt{R^2 - (t - t_0)^2} & |t - t_0| \leq R \\ 0 & \text{otherwise} \end{cases}, \quad (2)$$

where t_0 is the center of the profile with respect to t , R is the radius of the vessel, and α is a scaling factor that depends on the vessel contrast, detected x-ray fluence, and the imaging gain of the system. The background term $I_b(t)$ is generalized as

$$I_b(t) = \sum_{i=0}^N k_i \varphi_i(t), \quad (3)$$

where k_i are coefficients of orthonormal basis functions of polynomials $\varphi_i(t)$. In related literature^{22,23} and our own experience, a fifth order polynomial has been sufficient to model intensity variations from background anatomy and uneven x-ray exposure.

Including the model parameters explicitly, the noise-free modeled profile $I_0(t, \mathbf{a})$ is given by

$$I_0(t, \mathbf{a}) = I_b(t, \mathbf{a}) - (I_v(t, \mathbf{a}) * g(t)), \quad (4)$$

with $\mathbf{a}$ as the vector of model parameters $\{t_0, R, \alpha, k_0, k_1, \dots, k_N\}$. The maximum likelihood fit of the model to the observed profile (which may include noise) is found by optimizing $\mathbf{a}$ such that

$$[\mathbf{a}]_{\text{ML}} = \underset{[\mathbf{a}]}{\text{argmin}} \int [I(t) - I_0(t, \mathbf{a})]^2 dt. \quad (5)$$

When the processing is performed in the discrete domain, the discrete sum rather than a continuous integral is minimized. Figure 3 graphically illustrates the components of a model-based diameter measurement.

The line spread function $g(t)$ is not explicitly defined with this method; it must be determined by other means and inserted into the model described in Eq. (4). While the LSF can be measured independently, it is image plane and depth (Z) dependent and would require that the LSF be measured at the

FIG. 3. Vessel profile model. (a) The unblurred vessel model assumes a circular cross section with the profile shown in the figure. (b) The LSF is estimated based on actual measurements or an imaging model. (c) The unblurred profile is convolved with the LSF to create a blurred vessel profile. (d) The blurred profile (solid line) is subtracted from the background (dashed line) to approximate the observed profile (circles).

same position as the imaged vessel. In this study, the LSF is modeled at any position between the source and detector based on the gantry and x-ray source geometry, the position of the tomosynthetic image plane, the known image reconstruction parameters, an estimate of the focal spot size and shape, and an estimate of the vessel's 3D position as described in Sec. III A 4.

III. METHODS

III.A. QCA algorithm with SBDX

III.A.1. 2D vessel centerline

The sizing algorithm begins with the definition of a vessel centerline in a displayed SBDX angiographic image (multi-plane composite). An iterative method was implemented to define the centerline. A 2D piecewise cubic spline was fit through two to four user-selected points along the vessel centerline in the image. The remaining steps, described below, were automated in software. Intensity profiles perpendicular to the spline were sampled at 2 pixel intervals along the length of the spline. The model-based sizing algorithm, using an ideal (point) LSF, was applied to calculate the center of each vessel profile. Another smoothing spline was fit through the profile center points to create a more refined 2D centerline.

III.A.2. 3D localization of vessel borders

At 1 pixel intervals along the refined 2D vessel centerline, profiles were extracted perpendicular to the centerline and the model-based sizing algorithm was applied, again assuming an ideal LSF. Small ROIs were defined over the vessel edges along the profile for each sampled point as shown in Fig. 4. Given a point $\mathbf{x}$ along the 2D centerline and the normalized vector $\mathbf{c}$ perpendicular to the tangent of the center-

FIG. 4. ROIs for 3D vessel localization. (a) An intensity profile is extracted perpendicular to the centerline and two ROIs, $\text{ROI}^{(1)}$ and $\text{ROI}^{(2)}$, are defined at either edge of the vessel along the profile. (b) The ROIs defined on the composite image are applied to the score image stack. The 3D localization algorithm is executed on each ROI score stack to produce a Z coordinate for each ROI, $z_{\text{edge}}^{(1)}$ and $z_{\text{edge}}^{(2)}$. (c) The two Z coordinates from the ROI score stack correspond to opposite edges of the vessel. The two values are averaged to estimate $\bar{z}$, the Z coordinate of the 3D vessel centerline.

line at $\mathbf{x}$, two ROIs were centered at $(\mathbf{x} - \hat{R}\mathbf{c})$ and $(\mathbf{x} + \hat{R}\mathbf{c})$. ROI height and width were $2\hat{R}$, where $\hat{R}$ was the vessel radius estimate assuming an ideal LSF.

A “score image” stack was created from the single-plane image stack as described by Speidel *et al.*²¹ Each image was low pass filtered to reduce noise and vertical and horizontal gradient images of each smoothed image were created using vertical and horizontal 3×3 Prewitt²⁴ kernels. The absolute values of the gradient filtered images were summed together to make a score image. The score images from each plane position composed the score image stack $S(u, v, z)$, where u and v are the column and row pixel coordinates and z corresponds to the Z coordinate of the image plane. As shown in Fig. 2, the Z axis runs from the source to the detector, with $z=0$ at the source plane. In this case, z is only defined for discrete values corresponding to the Z coordinates of a finite set of single image planes. Using the pixel coordinates from a vessel edge ROI, the same ROI was extracted from each of the score images to create an ROI score stack $S_{\text{ROI}}(u, v, z)$. A maximum intensity projection $M_{\text{ROI}}(u, v)$ along the Z direction of $S_{\text{ROI}}(u, v, z)$ was calculated. $M_{\text{ROI}}(u, v)$ was then used to create a binary mask of the vessel edge $B_{\text{edge}}(u, v)$ by setting pixels in $M_{\text{ROI}}(u, v)$ with values greater than 80% of the maximum score in $M_{\text{ROI}}(u, v)$ to 1 and all other pixels to 0. The vessel edge score stack $S_{\text{edge}}(u, v, z)$ was defined as

$$S_{\text{edge}}(u, v, z) = B_{\text{edge}}(u, v) \cdot S_{\text{ROI}}(u, v, z). \quad (6)$$

Baseline subtraction of $S_{\text{edge}}(u, v, z)$ was performed by subtracting 40% of the maximum score in $M_{\text{ROI}}(u, v)$ from all values in $S_{\text{edge}}(u, v, z)$, followed by setting negative values to 0, which removed gradient influence from image noise and created $\hat{S}_{\text{edge}}(u, v, z)$. An overall Z coordinate for the vessel edge was calculated by summing the values of $\hat{S}_{\text{edge}}(u, v, z)$, such that

$$f_{\text{edge}}(z) = \sum_v \sum_u \hat{S}_{\text{edge}}(u, v, z). \quad (7)$$

The Z coordinate for the vessel edge z_{edge} was calculated as

FIG. 5. 3D vessel edge localization. (a) The vessel edge ROI was applied to the score image stack to produce the ROI score stack S_{ROI} . (b) M_{ROI} , the maximum intensity projection of S_{ROI} , was created. (c) A threshold equal to 80% of the maximum value in M_{ROI} is applied to M_{ROI} to create the binary mask of the vessel edge, B_{edge} . (d) The binary mask B_{edge} was applied to each image in the score image stack to create S_{edge} . A value equal to 40% of the maximum value in M_{ROI} was subtracted from each image in S_{edge} . Negative values were set to zero and the values in each score image were summed. (e) The Z position of the vessel edge in the ROI was calculated as the weighted average of the score sums, which was equivalent to the center of mass of the shown distribution.

$$z_{edge} = \frac{\sum_z f_{edge}(z) \cdot z}{\sum_z f_{edge}(z)}, \quad (8)$$

which was the “center of mass” of the score sums. Note that this assumes that the image plane set is nearly symmetrically distributed about the target vessel along the Z axis. Figure 5 shows each of the edge localization stages graphically. This same process was applied to the other score stack ROIs for the opposite vessel edge and the two Z coordinates were averaged [Fig. 4(c)]. The 2D coordinate (u, v) along the vessel centerline and the mean Z coordinate were converted to a 3D coordinate (x, y, z) and stored as a point along the 3D centerline of the vessel. This process was repeated for each of the profiles extracted along the 2D centerline to create a set of points along the 3D centerline of the vessel.

III.A.3. Vessel length measurements

A 3D piecewise cubic smoothing spline was fit through the series of 3D centerline points to reduce noise and create a continuous function. The physical length of the vessel segment corresponding to this 3D centerline was calculated by sampling the cubic spline at regular intervals and summing the arc length of a piecewise linear function through each of these points.

III.A.4. Vessel diameter measurements

Measurements of the physical diameter of a vessel, in mm, were performed using the model-based approach previously described with pixel-to-mm conversion factors and LSF approximations determined from the 3D vessel centerline. The 2D centerline in the composite image was sampled at 1.5 pixel intervals along its length and intensity profiles were extracted perpendicular to the centerline at each point. The 1.5 pixel spacing between adjacent profiles minimized the chance that the profiles would contain values extracted from the same pixels in the image, keeping the diameter measurements independent. Along the perpendicular profile, pixel values were sampled every 0.5 pixels using bilinear interpolation. For each sampled point along the 2D centerline and corresponding intensity profile, the pixel spacing and LSF were computed as described in the following paragraphs.

The sampled point along the 2D centerline was projected onto the 3D centerline and the closest image plane in the set of single-plane images to the projected point was identified. The pixel size p for this image plane was used to convert the width of the measured profile from pixels to physical dimensions (mm). Pixel size in a plane is given by the relationship

$$p = \frac{Z_d - z}{Z_d} \cdot \frac{\Delta_s}{m}, \quad (9)$$

where Z_d is the source-to-detector distance (1500 mm in the current prototype), z is the Z coordinate of the image plane, Δ_s is the spacing between adjacent focal spot positions in the scanning x-ray source (2.3 mm), and m is a reconstruction parameter set to 10 in this study. Figure 6 shows the pixel size p as a function of plane position.

The LSF component of the model fitting $g(t)$ was approximated based on the optical transfer function (OTF) model of the SBDX system.¹⁹ The OTF model for a single image plane reconstruction was dependent on x-ray source and detector geometry, reconstruction parameters, and the Z coordinate of the image plane position. In addition, the Z coordinate of the 3D centerline point relative to the image plane was incorporated into the OTF model to account for out-of-plane blurring effects. The OTF was assumed to be spatially invariant within the image plane. The 2D OTF was modeled as

$$\text{OTF}(\xi, \eta) = F(\xi, \eta) \cdot \text{sinc}(d\xi, d\eta) \cdot \text{sinc}(r\xi, r\eta) \cdot \text{sinc}(p\xi, p\eta) \cdot \text{sinc}(b\xi, b\eta), \quad (10)$$

where ξ and η are horizontal and vertical spatial frequencies and p is the pixel spacing for the image plane as previously described. The variable d is the detector element size scaled to the image plane, given by $d = \Delta_D(z/Z_D)$, where Δ_D is the detector element pitch. The parameter r is the size of the restoration kernel for the detector data and is equal to p times a constant. Out-of-plane blurring of a vessel region caused by the distance between the vessel and the closest image plane is modeled by convolution with a boxcar kernel of width b , such that

FIG. 6. Pixel size vs plane position. The pixel size of single-plane images decreases monotonically as the Z position of the plane increases. (Inset) Pixel size is a function of the position of the image plane (z), the source-to-detector distance (Z_d), the spacing between focal spot positions (Δz), and a constant m . Image field-of-view decreases as the image plane moves toward the detector and the number of pixels in each plane is fixed (e.g., 710×710 pixels).

$$b = |\Delta z| W_D / (Z_D - z - \Delta z), \quad (11)$$

where W_D is detector width and Δz is the signed distance of the vessel point to the image plane. All modeled components of the OTF with the exception of $F(\xi, \eta)$ represent rectangular elements. Hence, they are represented as sinc functions in the frequency domain.

The blurring due to the focal spot size $F(\xi, \eta)$ was modeled assuming a circular, pillbox shaped focal spot with unit area and is described by

$$F(\xi, \eta) = 2 \cdot \text{jinc}(\rho \sqrt{\xi^2 + \eta^2}), \quad (12)$$

where $\text{jinc}(X) = J_1(2\pi X) / (2\pi X)$, J_1 is a first order Bessel function of the first kind, and ρ is the physical radius of the focal spot scaled to the image plane. Focal spot size in the SBDX x-ray source is operator adjustable. For the source operating technique used in this study, the size of the focal spot at the source plane was measured to be approximately equivalent to a 0.9 mm diameter pillbox.

The LSF is dependent on the orientation of the extracted intensity profile within each image plane. The angle of the extracted profile relative to the image coordinate axes θ was calculated. According to the projection slice theorem, the LSF for the profile was the inverse Fourier transform of a line passing through the center of frequency space at angle θ relative to the spatial frequency axes, which is equivalent to

$$g_\theta(t) = \int_{-\infty}^{\infty} \text{OTF}(w \cos \theta, w \sin \theta) e^{2\pi i t w} dw. \quad (13)$$

The model-based algorithm was applied to each of the extracted intensity profiles using their corresponding pixel spacing and LSF. Model parameters (e.g., approximate diameter) were initialized using an intensity threshold method. The background intensity was initialized as the least-squares fit line going through the first and last five samples of the extracted profile, which were always outside of the vessel projection. Optimization of model parameters was performed using a Nelder–Mead simplex method. The results of the model fitting provided the physical diameter of the vessel (i.e., diameter = $2R$, where R is the model radius parameter after optimization).

III.B. Phantom validation

Vessel phantoms were used to validate the proposed algorithm. Vessel segments of different diameters were simulated using linear segments of solid MgCd alloy wire. This alloy is designed to have attenuation properties similar to iodinated contrast agent (350 mg/mL iodine) for diagnostic x-ray beam energies (60–120 kVp).²⁵ Wires with nominal diameters of 0.5, 0.7, 1.0, 1.4, 2.1, and 2.8 mm were used. The nominal length of each segment was 42 mm. Sheets of polymethylmethacrylate were used as the background attenuator and had a total thickness of 23 cm.

Different phantom configurations were used to test two hypotheses: (1) Vessel diameter and length can be accurately measured over a range of plane positions without the use of a magnification calibration step and (2) vessel diameter and length can be accurately measured when the vessel appears foreshortened in the 2D composite image. The first phantom configuration had vessels (one 4 cm length per diameter) oriented parallel to the image planes (i.e., all points along a vessel centerline at the same z -distance from the x-ray source). Images were acquired with the vessels positioned from 16 cm below to 16 cm above the mechanical isocenter of the gantry in 2 cm increments. In the second phantom, the vessel plane was placed on a stage that rotated around an axis through mechanical isocenter and orthogonal to the Z axis; rotating the vessel plane around this axis produced different degrees of vessel foreshortening in the 2D image display. Two 4 cm lengths per nominal diameter were imaged. Images were acquired with the vessel plane oriented 0° , 15° , 30° , 45° , 60° , and 75° with respect to the source plane. Figure 7 demonstrates the various appearances of vessel orientations in the SBDX multiplane composite images.

III.C. Imaging parameters

The vessel phantoms were imaged using the SBDX prototype hardware system.¹⁹ The vessel phantom was imaged at 15 frame/s, 100 kVp and 120 mA, using 71×71 focal spot positions. Each image plane had 710×710 pixels, with pixel width given by Eq. (9). For an isocenter plane image, the field-of-view was $11.4 \text{ cm} \times 11.4 \text{ cm}$ and the pixel size was $0.161 \text{ mm} \times 0.161 \text{ mm}$. The effective beam pulse width was 9.4 ms. The effective beam pulse width is the time between the first and last illumination of a given point in the object by the scanning beam during one frame period.

FIG. 7. Vessel phantom orientations. (a) To test the algorithm with different levels of vessel magnification, the phantoms were imaged parallel to the image plane from 16 cm above isocenter to 16 cm below [(b) and (c), respectively] in 2 cm increments. Note that the inverse geometry causes the magnification to increase as the object moves toward the detector. (d) To test the algorithm with foreshortened vessels, the vessel phantom plane was rotated about isocenter from 0° to 75° [(e) and (f), respectively] in 15° increments. The cross indicates isocenter. For clarity, the system dimensions have been exaggerated.

Single-plane images and multiplane composite images were reconstructed using a software simulation of the hardware-based reconstruction algorithm. For each acquisition, 16 single image planes spaced 12 mm apart were defined symmetrically above and below the Z position of the vessel phantom. The number of reconstructed planes matches the current hardware capabilities of the SBDX prototype and 12 mm plane spacing is sufficient to minimize the impact of out-of-plane blurring in the 2D composite image. Vessel measurements were performed on images integrated across 30 frame periods to test the algorithm under low noise conditions.

III.D. Error measurements

The error of diameter or length measurements of the vessel phantoms was calculated by subtracting the true dimensions from the measured dimensions produced by the device sizing algorithm for each phantom configuration. The true length and diameter were determined by measuring each vessel wire phantom with high precision calipers. Length measurement errors were averaged for segments with similar nominal diameters and the mean and standard deviation of length errors across all nominal diameters was calculated. The mean and standard deviation of diameter measurement errors were calculated over all diameter measurements for segments with the same nominal diameter in each configuration. Mean measurement errors represent systematic error (i.e., accuracy) and the standard deviation of the error represents the variability (i.e., precision).

To quantify changes in the apparent length of the vessel due to the phantom configuration, the length of the 2D centerline of each vessel (in pixel units) was calculated as a percentage of the reference 2D centerline length, which was

FIG. 8. Diameter error for different vessel magnifications. The mean diameter error for each vessel phantom diameter was calculated for vessels parallel to the image plane but at different distances from the source, producing different magnifications in the composite images. Average diameter errors remained relatively constant, remaining ≤ 0.15 mm for all diameters.

measured with the vessel phantom positioned at isocenter and parallel to the image plane (i.e., no foreshortening). The apparent full length of the vessel was calculated as the arc length of the 2D centerline after extrapolation to the ends of the vessel segment. Each “end” of the vessel phantom was defined as the point along the 2D centerline where the intensity was 50% of the maximum vessel contrast.

Length and diameter measurements were performed along the inner 75% of the 2D centerline of each segment. This avoids introducing measurement errors near the end points of the cylindrical segments, which have a changing cross sectional profile when angulated relative to an image plane.

IV. RESULTS

The length and diameter of each vessel phantom segment was successfully measured for each configuration. Diameter measurements on average were accurate to within 0.1 mm of the true diameter for all but the largest diameter vessels. Error variability increased for small diameter vessels (≤ 1.0 mm diameter). Typical length errors were less than 1.0 mm, with variability increasing as the apparent foreshortening of the vessel increased.

Imaging the vessel phantoms parallel to the image and source planes while varying the position in the Z direction produced images of nonforeshortened vessel wires at various magnifications. The apparent size (diameter and length) of the vessel phantoms was 13% smaller when positioned 16 cm below isocenter and 18% larger when positioned 16 cm above isocenter. Despite the changes in the apparent size of the vessel, the mean error of the vessel diameter measurements remained consistently less than 0.1 mm for all vessel diameters at all Z positions, except for the 2.8 mm diameter vessel measurements, which were within 0.15 mm (Fig. 8). As the apparent size of the vessel increased or decreased, the mean error of the diameter measurement remained consistent (Fig. 9). The precision of the diameter measurements was greater for larger diameter vessels. The variability of the measurements increased as the true diameter decreased, es-

FIG. 9. Measured diameter for different vessel magnifications. Plot (a) shows the measured diameter vs Z position for the 0.5, 1.0, and 2.1 mm diameter vessels. Plot (b) shows the measured diameter for 0.7, 1.4, and 2.8 mm diameter vessels. Dashed lines represent true diameter. Labels in right margin indicate the true diameter of each segment. Error bars represent ± 1 SD.

pecially for diameters ≤ 1.0 mm. Changes in the variability were irrespective of the apparent magnification of the vessel. The average number of diameter measurements per segment was 134 (range: 114–161).

As the angle of the vessels relative to the source plane increased (i.e., the vessel became foreshortened), mean diameter errors remained within 0.1 mm for smaller vessel diameters (0.5–1.4 mm) across all angles as well as for larger diameters (2.1 and 2.8 mm) at low to moderate angulation (0° – 45°). Errors exceeded 0.1 mm for larger diameters when foreshortening was severe (60° – 75°). There was a general trend for the mean error to increase as the angle of the vessels increased as shown in Fig. 10. Variability of the diameter measurements decreased as the angulation increased, as shown in Table I. The average number of diameter measurements per vessel segment was 97 (range: 34–139).

Length measurements of vessels parallel to the image plane imaged at different Z positions (i.e., different magnifications) were unaffected by the change in apparent size of the vessel. Figure 11 shows the measured length relative to the true length, averaged across all nominal diameters, compared to the apparent size of the vessels at each position along the Z axis. Across all vessel phantom positions and individual nominal diameters, each measured length was within 1% (~ 0.3 mm) of its true value.

FIG. 10. Measured diameter error for different angulations. The average diameter error showed little to no dependence on the vessel orientation for low to moderate foreshortening (vessel angle $\leq 45^\circ$). Diameter errors generally increased for all vessel diameters when the vessels were severely foreshortened (vessel angle $> 45^\circ$).

Length measurements of foreshortened vessels remained accurate despite large changes in apparent length. As the angle of the vessel relative to the image plane increased, the apparent length of the vessel decreased down to 28% of the nonforeshortened length. Averaged across all nominal diameters for a given orientation, the error was less than 1% (~ 0.3 mm). Length errors were greater for individual segments but remained less than 5% (~ 1.5 mm) overall. Variability in the length measurements tended to increase as the vessels appeared more foreshortened. Figure 12 shows the measured lengths relative to the true lengths compared to the changes in the apparent length of the vessels due to foreshortening.

V. DISCUSSION

Using a calibration-free QCA algorithm with tomosynthesis-based 3D localization, accurate diameter and length measurements were performed in a vessel phantom. The 3D localization method enabled accurate measurements over a 32 cm volume along the source-detector axis. The vessel contrast and diameters used in this study were representative of those encountered in coronary angiography with arterial injection of iodinated contrast. In orientations where the vessels appeared foreshortened in the images, the algorithm was able to calculate accurate diameters and lengths.

Interventional devices (e.g., balloons and stents) are available in a variety of diameters and lengths in order to closely match lesion length and the adjacent reference (healthy) vessel diameter. Typical device lengths are available in 4–5 mm increments and diameters are available in increments of 0.25 mm with the narrowest devices having diameters of ~ 2.0 mm. Based on our experiments, the errors in the estimated diameters and lengths are within these incremental values for vessel diameters that can be treated with interventional devices (≥ 2.0 mm). This suggests the potential for the algorithm to provide guidance during clinical device size selection. While accurately measuring narrow vessels (< 2.0 mm) has diagnostic value (e.g., for measuring %

TABLE I. Diameter error with apparent foreshortening. Angle corresponds to angle between the vessel center axis and the image planes.

Nominal diameter (mm)	Mean error ± 1 SD (mm)					
	0°	15°	30°	45°	60°	75°
0.5	-0.04 ± 0.30	-0.08 ± 0.26	-0.03 ± 0.25	0.02 ± 0.23	0.06 ± 0.20	0.05 ± 0.15
0.7	-0.08 ± 0.22	-0.06 ± 0.19	-0.06 ± 0.21	-0.03 ± 0.17	0.03 ± 0.13	0.04 ± 0.07
1.0	-0.03 ± 0.07	-0.03 ± 0.07	-0.02 ± 0.07	-0.02 ± 0.06	0.01 ± 0.05	0.05 ± 0.04
1.4	0.00 ± 0.06	0.00 ± 0.05	0.01 ± 0.05	0.01 ± 0.04	0.02 ± 0.04	0.06 ± 0.03
2.1	-0.01 ± 0.04	0.00 ± 0.04	0.00 ± 0.04	0.02 ± 0.03	0.03 ± 0.03	0.11 ± 0.03
2.8	0.03 ± 0.04	0.05 ± 0.05	0.05 ± 0.04	0.07 ± 0.04	0.11 ± 0.04	0.20 ± 0.05
Overall	-0.02 ± 0.16	-0.02 ± 0.14	-0.01 ± 0.14	0.01 ± 0.13	0.04 ± 0.11	0.09 ± 0.09

stenosis), it is less critical for device sizing because devices that can match these small diameters are not available.

One noticeable trend when measuring the vessel phantoms in cases of severe foreshortening was the increased overestimation of the vessel diameter. The assumption that the LSF was constant over the Z extent of the vessel cross section appeared to be adequate for shallow foreshortening angles but not for steep angles. The clinical impact of the observed overestimation of severely foreshortened vessels is likely minimal because trained interventionalists recognize extreme foreshortening effects and are accustomed to adjusting the projection angle to avoid this problem. Under conditions where the vessel angle is less than 60° relative to the image plane, the diameter and length are accurately measured using the proposed algorithm.

Interestingly, the same severe phantom orientation that increased the mean diameter error resulted in a lower variation of the diameter measurements, specifically for smaller diameter vessels. As the vessels changed orientation from parallel to the source plane to being angled in the direction of the x-ray beam, the path length through the vessel phantom increased. This led to greater attenuation and improved the

contrast to noise ratio (CNR), thereby lowering the random variation of the diameter measurements. However, for a given orientation, smaller diameter vessels will have a lower CNR than larger vessels and diameter measurements of smaller diameter vessels will correspondingly have lower precision.

All absolute dimensions were quantified without requiring a separate calibration step. Various calibration methods have been proposed for conventional QCA algorithms,¹⁸ the most common of which is to image a contrast filled catheter near the vessel of interest. Alternatively, an external calibration object, such as an x-ray visible ruler with more conspicuous features can be used; however, this increases calibration uncertainty because a vessel within the body and a calibration object outside of the body experience different magnifications. Eliminating the calibration step altogether removes a potential source of error and improves procedure workflow, which increases the likelihood that such a QCA method would be used clinically.

Recent 3D coronary vessel reconstruction methods may also provide alternatives to conventional vessel measurement methods. Multiple angiographic views or rotational angiography acquisitions are typically used as the source images. These methods may use *a priori* assumptions to model coronaries,^{26–29} while others use volumetric reconstruction techniques.^{30,31} In terms of device sizing, 3D reconstructions may be used to determine optimal angiographic views that minimize foreshortening and overlap,³² which may improve conventional QCA results. Vessel bifurcation angles and tortuosity can also be quantified from 3D reconstructions, which may help support interventional decision making. The feasibility of measuring vessel segment length and diameter directly from these 3D reconstructions has been demonstrated in phantoms and *in vivo*.³¹ Although 3D reconstructions may provide other useful information in addition to vessel dimensions, they require image acquisition from multiple gantry angles, whereas the proposed SBDX QCA method requires only a single gantry angle.

The benefit of 3D vessel localization is the determination of vessel magnification to convert pixel measurements to physical dimensions, which is necessary for any QCA technique. Either derivative-based or model-based edge detection

FIG. 11. Measured relative length for different vessel magnifications. The measured relative vessel length (measured vessel length/true vessel length) (gray) was calculated and averaged for all vessel segments at each Z position. Apparent length (black) is the relative length measurement expected using conventional methods, assuming the magnification calibration was performed only at isocenter. Error bars represent ± 1 SD.

FIG. 12. Measured relative length for different angulations. Relative vessel length (measured vessel length/true vessel length) (gray) was averaged for all vessel segments imaged at each orientation. The apparent vessel length (black) assumes a conventional magnification calibration and no compensation for the degree of vessel foreshortening. Error bars represent ± 1 SD.

methods could have been used in this investigation. The model-based approach was chosen because it allowed us to also exploit our knowledge of the z -dependent LSF in any image plane using the OTF model. This may produce more accurate results at small vessel diameters (<1.0 mm) where the apparent diameter is mainly due to the system LSF. Further work is required to determine which diameter measurement method produces the best results within our calibration-free QCA framework.

This study has several limitations that require future work. Although the algorithm produced satisfactory results, the validation was performed on straight vessel phantoms imaged under low noise imaging conditions with a flat background. The results thus represent the limit against which future results obtained under clinical conditions (background anatomy, branching vessels, eccentric lesions, and higher noise levels) can be compared. In this initial study, vessel analysis was performed offline in software. A fast intraprocedural implementation would be desirable. Minimizing the time required to perform the measurements will increase the likelihood that interventionalists will use the algorithm to guide clinical decisions.

VI. CONCLUSIONS

Using inverse geometry x-ray fluoroscopy, we have demonstrated the feasibility of using tomosynthesis imaging to perform calibration-free measurements of vessel dimensions in phantoms. Accurate measurements of diameter and length were obtained using a wide range of foreshortening angles and vessel positions along the source-detector axis. Reducing the overhead of the calibration step simplifies the QCA measurement, making it easier to use in an interventional setting. This could lead to better device selection, thereby improving outcomes and reducing costs.

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